Interactive Effects of Load Carriage Magnitude and Marching Velocity on Kinetic, Kinematic and Spatiotemporal Gait Characteristics in Recruit-Aged, Physically Active Females

by

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BS Exercise Science, University of Pittsburgh, 2016

Submitted to the Graduate Faculty of

School of Health and Rehabilitation Sciences in partial fulfillment

of the requirements for the degree of

Master of Science

University of Pittsburgh

2019

UNIVERSITY OF PITTSBURGH

SCHOOL OF HEALTH AND REHABILITATION SCIENCES

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Marching with heavy load is a large component of military training and standard operations. The rate of the march may approach or surpass the velocity at which warfighters would prefer to run, however, they are strongly encouraged to maintain a walking gait. Previous research has observed gender differences in biomechanics and prevalence of injury during load carriage. The primary purpose of this study was to investigate the effects of load carriage magnitude on lower limb kinematics, kinetics, and spatiotemporal metrics at velocities around the gait transition point (GTP) in female subjects. As a secondary aim, the effects of load carriage on coordination variability were assessed. Twelve recreationally active females (24.75 ± 2.26 years) completed three testing sessions. 3D biomechanics were captured via 12 infrared cameras (Vicon Motion Systems, Oxford, UK) and a Bertec fully instrumented treadmill (Bertec Corporation, Columbus, Ohio). Prior to data collection, GTP velocity was determined by averaging 3 walk-to-run trials for each load condition. Experimental trials were conducted at body weight (BW), and with loads; +25%BW, and +45%BW. For each load condition, participants walked (WK), ran (RN) or force-marched (FM) at velocities around their GTP velocity. Kinematic data were acquired at 100Hz and kinetic data were acquired at 1000Hz. Repeated measures analysis of variance were conducted to examine the effect of load and marching velocity on sagittal plane kinematics, kinetics, and spatiotemporal metrics. Results showed that increases in load carriage magnitude increased stride width, stance time, and double support time and decreased stride length and flight time. Knee flexion and ankle dorsiflexion at mid-stance increased with increases in load. Overall, joints were more flexed at each gait event during running. Lower extremity relative joint moments decreased with increases in load. Hip and knee moments were greater during forced-marching at heel-strike and toe-off. All joint moments were greater during running at mid-stance. Increases in load did not appear to decrease the variability of shank-thigh coupling. Additionally, there was no apparent difference in coordination variability between RN and FM trials. Future studies should investigate the biomechanical differences between trained and untrained males and females.

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Preface

I am grateful for the opportunity that I was provided in taking on this research project. There are a number of people who I would like to thank for their contributions to this intricate project. First and foremost, I would like to thank my thesis committee, Dr. Katelyn Allison, Dr. Shawn Flanagan, Dr. Mita Lovalekar, Dr. William Anderst, and my committee chair, Dr. Chris Connaboy. This project would not have been possible without all their efforts, encouragement, and faith in me along the way.

Thank you to all the Neuromuscular Research Laboratory graduate students and faculty and staff. From engaging in conversation, exchanging thoughts, and in some cases, even freeing up my time with other work commitments so that I could focus on this project. Each person in the lab has helped me, in one way or another, throughout this process.

I would also like to thank the faculty and staff at the Biodynamics Laboratory for all the support and for allowing us to use their space and equipment to conduct this study. It was a privilege to collaborate with such a dedicated and knowledgeable group of researchers.

Lastly, I would like to thank Dr. Freddie Fu for funding this research project through the "Freddie H. Fu, MD Graduate Research Award".

1.0 Introduction

Military personnel are required to traverse extensive distances on foot while carrying and wearing heavy loads. Loads carried by the warfighter have increased substantially over the decades and marching with heavy loads has become a large component of training and standard operations(Dean, 2008; Drain, Orr, Attwells, & Billing, 2012; J. Knapik, Reynolds, Santee, Friedl, & Borden, 2010; J. J. Knapik, Reynolds, & Harman, 2004). With repeated bouts of load carriage, the warfighter incurs a greater risk for traumatic and overuse injuries during both acute exposures(Michael LaFiandra & Harman, 2004; Sell et al., 2010) and prolonged exposures(Attwells, Birrell, Hooper, & Mansfield, 2006; Drain et al., 2012; E. A. Harman, Han, Frykman, & Pandorf, 2000; J. Knapik, Reynolds, Staab, Vogel, & Jones, 1992). Marching with a backpack has been shown to alter certain physiological(Abe, Muraki, & Yasukouchi, 2008; Abe, Yanagawa, & Niihata, 2004; Quesada, Mengelkoch, Hale, & Simon, 2000; Soule, Pandolf, & Goldman, 1978; Stuempfle, Drury, & Wilson, 2004; Treloar & Billing, 2011) and biomechanical(Attwells et al., 2006; S. A. Birrell & Haslam, 2009; Stewart A. Birrell & Haslam, 2010; Goh, Thambyah, & Bose, 1998; E. Harman, Hoon, Frykman, & Pandorf, 2000; E. A. Harman, Han, & Frykman, 2001; E. A. Harman et al., 2000; Kinoshita, 1985; Tilbury-Davis & Hooper, 1999) aspects of gait. Alterations in ground reaction forces, lower limb kinematics, and spatiotemporal parameters of gait as a result of load carriage place a great deal of stress on the musculoskeletal system.(Caron, Lewis, Saltzman, Wagenaar, & Holt, 2015; E. A. Harman et al., 2001; J. Knapik et al., 2010; J. J. Knapik & Reynolds, 2016) While the musculoskeletal system is fully capable of withstanding sizable loads and enduring a considerable amount of stress, the amplitude and frequency at which load carriage tasks are conducted during military training and operations is a cause for concern.(Dean, 2008) Load carriage for extended periods of time has been shown to negatively impact marksmanship and grenade throwing, which are critical tasks for infantry personnel and other supporting elements on the front lines during combat operations.(J. Knapik et al., 1991) Decrements in the performance of these tasks may compromise the warfighter and his or her comrades.

Compounding on the biomechanical stress of the load carriage mass, marching velocity will also have marked effects on both kinematics and kinetics during a load carriage task. The velocity at which most individuals transition from walking to running, also known as the Gait Transition Point (GTP), occurs spontaneously at approximately 2.09 m/s.(Segers, 2006) The addition of a load may alter this preferred transition velocity.(Keren, Epstein, Magazanik, & Sohar, 1981; J. Knapik, Harman, & Reynolds, 1996) Marching velocity is heavily dependent on the mission requirements and may approach or even surpass the unloaded GTP. However, warfighters are strongly encouraged to maintain a walking gait rather than transition to running. Little is known about the effect of load carriage mass on the GTP, with only a few studies examining the impact of additional load on the GTP.(Brown, O'Donovan, Hasselquist, Corner, & Schiffman, 2014; F. J. Diedrich & W. H. Warren, 1998) Further research is required to investigate this imposed walking, or forced-march, undertaken by warfighters, as it may have significant biomechanical implications since warfighters are forced to adapt a less optimal and potentially deleterious gait pattern.

The more recent inclusion of females in combat centric roles warrants research to investigate the effects of load carriage on their biomechanics. Prescribed, mission specific, combat loads are typically absolute, regardless of the warfighters stature or body mass. Knowing the sex differences in anatomy and physiology, it’s reasonable to postulate that females will adopt different biomechanical strategies to manage these ‘absolute’ combat loads. There is an abundance of research on the independent effects of load carriage mass and marching velocity on physiological characteristics and a fair amount of research on the biomechanics, a majority of which has been conducted on male subjects. However, very few studies have investigated the interaction effects of load carriage mass and marching velocity on biomechanics and no previous studies have examined the interaction effects in females, and specifically their interaction at, and around the GTP. Furthermore, there is a gap in our understanding of the motor control and coordination of movement during load carriage. Deficits in coordination have been linked to increased risk of injury in athletic populations. Coordination and motor control are largely unexplored topics in load carriage research and much has yet to be established on the understanding of exactly what impact load carriage has on coordination. More research is needed to explore the interaction effects of load carriage mass and marching velocity on biomechanics, especially in a sample of female subjects. Additionally, research on the coordination and control of load carriage is scarce and investigating the stability/instability of gait coordination patterns under load at a preferred-stride and a forced-march may identify any suboptimal patterns.

1.1 Load Carriage in the Military

1.1.1 Historical Perspective

Throughout history, equipment carried by the warfighter has served four salient functions: sustainment (i.e. food and water), protection (i.e. body armor), lethality (i.e. weapon systems and ammunition), and command and control (i.e. communications equipment).(Drain et al., 2012) Prior to the 18th century, early military units used auxiliary transport (i.e. horses, carts, assistants) to carry the majority of their equipment.(Lothian, 1921) Equipment hauled by auxiliary transport primarily consisted of weapons, protection, and provisions.(Lothian, 1921) On the march, warfighters seldom carried loads greater than 15kg.(Lothian, 1921) Warfighters bore lighter loads on the march to preserve their energy for bearing heavier loads during combat. From the 18^{th} century on, the use of auxiliary transport was reduced and consequently the load carried by the warfighter increased.(Lothian, 1921) The burden of transporting mission essential equipment was placed upon the troops. Troops began carrying heavier loads during approach marches and lighter loads when actively engaging the enemy.

Prior to the second world war, efforts to formally study load carriage in the United States were insubstantial. WWII led to a number of situations in which the mission dictated warfighters to transport heavy loads for long distances. Drawing from those experiences throughout the war, several studies were conducted by the US Army Field Board No.3 shortly after the war ended. The board studied loads carried by individual positions and made suggestions for decreasing the load to increase combat effectiveness. They recommended that riflemen carry 18kg in the worst conditions and a maximum of 25kg on the march.(Bailey & McDermott, 1952) Just over a decade later, the US Army Combat Developments Command supported these load recommendations and devised the notion of load echeloning which defined loads based upon the condition.(ACDC, 1962, 1964) In 1987, the US Army Development and Employment Agency adopted and expanded the concept of load echeloning. The load carried by the warfighter was termed the *combat load* and defined all required equipment that is essential to mission accomplishment.(ADEA, 1987) Combat load was further categorized by *fighting load* and *approach march load*. Fighting load defined the load carried when conflict was expected or when conducting covert operations and included the warfighters clothing, load-bearing equipment, helmet, weapons, ammunition, and rations.(ADEA,

1987) Approach march load defined the load carried in prolonged operations and included the combat load plus a pack loaded with a sleeping roll, extra clothing, extra ammunition, and extra rations.(ADEA, 1987)

The most current US Army doctrine on load carriage recommends no more than 22kg (or 30% BW) for fighting loads and 33kg (45% BW) for approach march loads.(Army, 1990) However, with the ever-faster advancements in technology and armament, improvements in ballistic protection, and the complexity and unpredictability of modern warfare, warfighters are continuously burdened with heavier loads. To date, it is not uncommon for foot marches to be conducted with loads nearing or exceeding 50% of the individuals' body weight.(Dean, 2008; J. J. Knapik et al., 2004) A survey of combat loads carried by U.S. troops of various occupations during recent engagements reported fighting loads, approach march loads, and emergency approach march loads averaging 29kg (35% BW), 46kg (57% BW), and 60kg (78% BW), respectively.(Dean, 2008) Emergency approach march load defines loads carried through terrain that is impassable by vehicle or where ground/air transportation is not available.(Dean, 2008) These circumstances would require warfighters to carry all equipment essential to their respective occupations and may span several days over distances of 20km a day.(Dean, 2008) Some of the most substantial emergency approach march loads were seen in Mortar Section Leaders and Fire Support Officers who carried loads averaging 68kg (91% BW) and 65kg (99% BW), respectively.(Dean, 2008) The findings from this survey clearly indicate that the modern warfighters combat load considerably exceeds the recommendations establish by US Army doctrine. Heavy loads carried for long durations have been shown to alter several biomechanical characteristics, placing a great deal of stress on the musculoskeletal system. With repetition, the compounding stress of excessive loading and unfavorable biomechanical adaptations may predispose the warfighter to overuse musculoskeletal injuries.

1.1.2 Associated Injuries

Injuries most commonly reported with load carriage tasks include foot blisters, metatarsalgia, stress fractures, knee pain, low-back injuries, rucksack palsy, and local discomfort/fatigue.(J. Knapik et al., 1996; J. J. Knapik et al., 2004) An association between load carriage and the aforementioned injuries is evident, however, a direct link between the two has yet to be established. There is a scarcity of research investigating load carriage injury mechanisms. However, increases in load have been shown to alter gait mechanics(Attwells et al., 2006; E. Harman et al., 2000; E. A. Harman et al., 2001) and increase energy expenditure(J. Knapik et al., 2010; Patton, Kaszuba, Mello, & Reynolds, 1991; Quesada et al., 2000; Soule et al., 1978), musculoskeletal stress(Stewart A. Birrell & Haslam, 2010; Caron et al., 2015; Goh et al., 1998; Michael LaFiandra & Harman, 2004), and rate of fatigue.(Fallowfield, Blacker, Willems, Davey, & Layden, 2012; Grenier et al., 2012) These factors may increase the risk of musculoskeletal injuries. Although some of the more acute injuries may seem negligible (i.e. foot blisters, discomfort/fatigue), they are no less worth discussing as they may have a negative impact on the mobility and maneuverability of the warfighter. The emergence of an antalgic gait resulting from foot blisters or fatigue may place unaccustomed stress on certain musculoskeletal structures.(J. J. Knapik et al., 2001) What's more, sustained acute injuries that are not provided adequate time to properly heal may immutably alter biomechanics and predispose the warfighter to other, potentially more grievous, musculoskeletal injuries.(Bush, Brodine, & Shaffer, 2000)

Among all musculoskeletal injuries reported in the United States Armed Forces, injuries of the back and lower extremities are the most common injury sites.(Hauret, Jones, Bullock, Canham-Chervak, & Canada, 2010; Robin Marc Orr, Pope, Johnston, & Coyle, 2014) Moreover, 82.3% of all back and lower extremity musculoskeletal injuries were reported to be a result of cumulative trauma, or overuse injury.(Hauret et al., 2010) Studies specifically investigating load carriage injuries fall into two main categories. The first category comprises those studies investigating the incidence of injury from a single load carriage excursion. Two different studies reporting injury incidence calculated incidences of 24%(J. Knapik et al., 1992) and 90%.(Dalén, Nilsson, & Thorstensson, 1978) The large variation observed between studies may be attributed to the difference in prescribed loads and distances. However, the majority of injuries reported for both studies involved either the back or lower extremities.(Dalén et al., 1978; J. Knapik et al., 1992) The second category involves studies investigating a single load carriage related injury to identify risk factors and outline injury prevention strategies.

1.1.3 Risk Factors for Load Carriage Related Injuries

The risk factors for injuries associated with load carriage are similar to risk factors for other training related injuries(J. Knapik et al., 1996). These risk factors can be grouped into one of two categories; modifiable or non-modifiable. Modifiable risk factors are those factors that can be controlled or altered by the individual. Modifiable risk factors include.(B. H. Jones, 1983; J. Knapik et al., 1996)

1. Heavy load
2. Improper load distribution
3. Long marching distance

4. Improper fitting or old footwear
5. Low Aerobic fitness
6. High BMI
7. Lower quadriceps flexibility
8. Lower vertical jump performance
9. Greater hip adduction
10. Lower isometric strength in hip adduction, knee extension, and knee flexion

Conversely, Non-modifiable risk factors are those factors that cannot be manipulated. Non-modifiable risk factors include.(B. H. Jones, 1983; J. Knapik et al., 1996)

1. Old Age
2. White ethnicity
3. Female gender
4. Terrain
5. Tall stature
6. Fatigue
7. Greater navicular drop
8. Greater patellar mobility
9. Previous injury

The modifiable risk factors for load carriage related injuries are among the most commonly implicated factors in load carriage research. They are perhaps the most important factors as each is amenable to intervention for the purpose of injury prevention.

1.2 Biomechanical Aspects of Load Carriage

1.2.1 Load Carriage Mass and Marching Velocity

When assessed independently, load carriage mass and marching velocity each have distinct effects on posture and gait. With the addition of an external load, the most prominent biomechanical adaptation observed is an increased forward lean of the truck to shift the pack center of mass (CoM) over the feet. Heavy backpack loads and positioning the load low within the pack have been shown to exacerbate this forward lean.(Bobet & Norman, 1984; Martin & Nelson, 1986) The increased forward rotation about the hip or ankles tends to shift the body's CoM further over the front half of the foot, which may increase the likelihood of foot injuries. In general, at a fixed pace, stride length and swing time decrease, and stride frequency and double-support time increase with increased load.(Attwells et al., 2006; Kinoshita, 1985; Martin & Nelson, 1986) However, there is contradictory evidence as to whether most individual shorten their stride length.(E. Harman et al., 1992) With respect to kinetics, heavy backpack loads significantly increase ground reaction forces (GRF), breaking forces at heel strike, propulsive forces at toe off, lateral forces during stance phase.(E. Harman et al., 1992; Kinoshita, 1985; Majumdar, Pal, & Majumdar, 2010) Increased moments about the trunk, hips, and knees are also observed with the addition of load.(Attwells et al., 2006; Brown et al., 2014) Regarding lower extremity kinematics, load carriage increases range of motion (ROM) at the hips, knees and ankles. The increase in knee flexion at heel strike is suggested to be a shock absorption mechanism(Attwells et al., 2006; Kinoshita, 1985), although one study did not observe such phenomena in those with more load carriage experience.(Tilbury-Davis & Hooper, 1999) The external load and resultant biomechanical adaptations place a considerable amount of stress on the musculoskeletal system, specifically the vertebral column and lower extremities. In training and operations, the warfighter is forced to maintain pace, often well past the onset of muscular fatigue, an identified risk factor for injury.

At a fixed load with increases in marching velocity, stride length and stride frequency increase while swing time and double-support time decrease. Under the same conditions, increases in GRF at both heel-strike and toe-off are observed with decreases in force during the mid-stance. As marching velocity increases, so does the ROM of the hips, knees, and ankles. The body lowers its CoM by increasing maximum hip and knee flexion and magnifying ankle dorsiflexion. Increases in hip and knee flexion are believed to emerge as a protective mechanism, absorbing some of the impact forces. Basic patterns of moments around the ankles, knees, and hips can be observed, although individual differences have been noted with the largest deviations occurring around hip and knee moments. Larger hip extensor and knee flexor impulses are seen in some individuals during running and sprinting. It is believed that this pattern emerges to minimize horizontal breaking force and improve running efficiency. However, there is evidence linking these larger impulses to hamstring injury.(Mann & Hagy, 1980)

While understanding the independent effects of load carriage mass and marching velocity are important, understanding how these variables interact to effect biomechanics is equally as important. Warfighters must be prepared to carry various loads at various marching velocities, in some scenarios, within a single mission. The mission objectives will dictate what and how much gear and armament are necessary and the timetable for the operation. It is highly unlikely that either variable, load or velocity, will remain constant throughout the entire mission. Only a couple of studies have investigated the effects of both load and velocity on biomechanics.(Brown et al., 2014; E. A. Harman et al., 2001) The results showed that, for the most part, the effects of increased marching velocity were the same regardless of backpack load and vice versa, with the exception of a few variables. Stride frequency, stride time and downward impulses all exhibited a statistical interaction, meaning that the effects of increased marching velocity were not the same for all backpack loads and vice versa. The evidence strongly indicates that the combination of fast marching velocities and heavy loads present a relatively high level of risk for fatigue and injury. Although these studies reveal important details and provide some insight on load-velocity interaction effects, none of them examined these effects at velocities around the GTP in female subjects.

1.2.2 Gait Transition

Walking and running each have their own unique biomechanical characteristics with very distinct fluctuating patterns of potential energy and kinetic energy. Walking is characterized by an out-of-phase fluctuation of potential energy and kinetic energy.(Segers, 2006) Kinetic energy is converted to potential energy as the foot is lifted to take a step and vice versa as the foot is returned to the floor. Walking is often described as an "inverted pendulum", with the body's center of mass moving in an arc shape upon a rigid stance leg. Walking is also characterized by the presence of a double support phase, a point in time within the gait cycle where the body is supported by both legs. Duty-factor is defined as the fraction of the stride for which each foot remains on the ground and is used to quantitatively differentiate walking from running. Walking is characterized by a duty-factor of greater than 0.5.(Segers, 2006) Alternatively, running is characterized by an in-phase organization of potential energy and kinetic energy.(Segers, 2006) The fluctuations of potential and kinetic energy share a more direct correlation and a considerable amount of energy is recovered through storage and release of energy from the elastic properties of the various muscles, tendons, and ligaments acting to carry out the movement. The dynamics of running are often compared to a bouncing ball. A flight phase is observed in running where both feet break contact with the ground and the body is suspended in air for a few milliseconds. Running is characterized by a duty-factor of less than 0.5.(Segers, 2006) Although walking and running are two easily discernable patterns of locomotion(Segers, 2006), the understanding of the transition point between these gait patterns is not as straightforward. Furthermore, the effects of load carriage on gait transition is a fairly unexplored topic.

The increase of walking velocity beyond a critical velocity will evoke a walk-to-run transition (WRT). Conversely, the decrease of velocity below a critical velocity will generally evoke a run-to-walk transition (RWT). Stride length and stride frequency adjust to changes in velocity (and load) to maintain an optimal gait pattern, or an optimal combination of stride length and frequency.(Attwells et al., 2006; Kinoshita, 1985; Martin & Nelson, 1986; Segers, 2006) It has been found that this optimal combination emerges to reduce energy expenditure.(Cavanagh & Williams, 1982; Mercier et al., 1994) The GTP occurs spontaneously at a velocity of approximately 2 m/s (for both WRT and RWT) although walking can be maintained at higher speeds.(Segers, 2006) The significance of the GTP as a tangible event is often overlooked in research.(Segers, 2006) When focusing solely on the spatiotemporal definitions of gait (duty-factor), humans are either walking or running, meaning a presence of either a double stance phase or flight phase, respectively. From this viewpoint, gait transition occurs very abruptly and seamlessly in one stride. However, an increase in variability of intralimb coordination, described as a non-equilibrium phase, can be observed leading up to the GTP.(F. J. Diedrich & W. Warren, 1998) It is widely accepted that the transition likely occurs in one or two strides but stages of preparation and post-transition adjustment appear to be evident upon closer examination.(F. J. Diedrich & W. Warren, 1998) While the exact determinants or triggers of gait transition have been argued upon, the consensus is that energy optimization is the supreme driver.(Diedrich & Warren, 1995; Mercier et al., 1994)

An Increase in load carriage mass has been negatively correlated with self-selected marching velocity.(Hughes & Goldman, 1970; J. Knapik et al., 1996; J. Knapik, Johnson, Ang, Meiselman, & Bensel, 1993; J. J. Knapik et al., 1997) Moreover, increases in load carriage mass were associated with decreases in GTP velocity in a sample of physically active civilian males.(Keren et al., 1981) Thus, as an individual is equipped with heavier loads, their self-selected marching velocity and GTP may decrease in a linear manner.(J. J. Knapik et al., 1997) Knowing that changes in load and velocity have considerable effects on the biomechanics of gait, it is imperative to closely monitor and promptly investigate the impact of any increase in demand. Understanding what effect modern military load carriage demands have on the biomechanics of the warfighter is important for informing military leaders and reshaping standard procedures to safely and effectively meet current demands. As previously discussed, a destabilization in coordination can be seen in periods just before and after gait transition. Forced-stride marching, at a velocity beyond that at which the GTP would normally occur, may inadvertently promote the maintenance of this destabilized gait pattern. Previous research has found an association between poor coordination and non-contact musculoskeletal injuries.(Swanik, 2015) Increased muscular stress resulting from heavy load carriage combined with maintenance of an unfavorable gait pattern at high marching velocities may put the warfighter at a greater risk for sustaining a musculoskeletal injury. Assessing what impact load carriage has on this non-equilibrium phase and the differences, if any exist, between coordination patterns of forced-march and preferred-stride are important for informing current military doctrine.

1.2.3 Coordination and Control

Walking and running are fundamental tasks that we begin exploring and mastering early on in child development. While these tasks may seem simple from a macroscopic perspective, they are in fact very complex movement patterns involving the coordination of multiple joints and muscle groups that are governed by a hierarchical order of coordinative structures. Coordination of movement was previously understood to be controlled entirely by the central nervous system.(Turvey, 1990) This cognitive theory attributes the control and coordination of movement to a plethora of motor programs and schemas coded to govern all aspects of the desired movement.(Turvey, 1990) However, Bernstein's "degrees of freedom problem" highlighted the infinitely complex nature of the control of coordinated movement.(Turvey, 1990) Bernstein postulated that given the immense number degrees of freedom and the complexity of human movement, it was unlikely that a motor program could account for all possible degrees of freedom within a movement system.(Turvey, 1990) These revelations created a divergence from the top-down model proposed by a cognitive theory of movement control.

The dynamical systems theory portrays human movement as a highly complex network of codependent subsystems that work concurrently to execute a desired movement pattern.(Turvey, 1990) Interaction between subsystems and exploration of the organismic, task, and environmental constraints for a goal-directed action drives the evolution of coordinated movement. Self-organization is a term long used in other physical and chemical sciences to describe the spontaneous emergence of an organized pattern of activity within a system.(Turvey, 1990) When applied to the dynamical systems theory for coordination and control of human movement, self-organization describes the tendency for multiple subsystems to collectively establish a synergistic relationship. Through the collective organization of these subsystems and all their available degrees of freedom, a single functional unit designed to carry out a specific task emerges.(Turvey, 1990) This framework presents a plausible solution to Bernstein's degrees of freedom problem. Applying the concept of self-organization to the dynamical systems theory provides a logical understanding for the development and description of coordination.

Using the framework of self-organization and the dynamical systems theory, the behavior of a system can be described by, and put in terms of a 'collective' or 'order' parameter and a 'control' parameter.(Turvey, 1990) The 'collective' or 'order' parameter refers to the variables used to describe the organizational status of a system (i.e. kinetics, kinematics, spatiotemporal characteristics).(Turvey, 1990) The 'control' parameter (i.e. marching velocity, load carriage mass) is a variable that drives the evolution of an order parameter (i.e. stride frequency, stride length).(Turvey, 1990) As a control parameter is manipulated (increased or decreased), it will likely alter the stability (increased or decreased variability) of the order parameter in effect. For example, a warfighter carrying a ruck is able to maintain a walking gait over a wide range of velocities. As marching velocity increases (control parameter), the order parameter begins to shift away from the preferred walking gait and the coordination pattern for walking becomes less stable. Eventually, a critical velocity is reached where the warfighter transitions to running, reorganizing the system and reestablishing stability. In short, a transition occurs when the control parameter reaches a critical value and the order parameter shifts from one stable attractor (walking) to another (running).

Coordination and control of gait during walking and running is well-studied(F. J. Diedrich & W. Warren, 1998; Diedrich & Warren, 1995), however it is a comparatively unexplored topic in the scope of load carriage research. Coordination and control of a movement pattern is essential to ensure that the system operates within its envelope of function and below the threshold of bone and tissue deformity. A great deal of evidence implicates uncoordinated movement and poor movement quality in non-contact musculoskeletal injuries.(Swanik, 2015) A coordinated movement pattern emerges as a result of ample movement exploration and subsequent motor learning. With adequate load carriage training that incorporates the fundamental training principles of progression and overload, the warfighter will likely establish optimal, coordinated gait patterns across a wide range of loads and marching velocities. While previous studies have attempted to define the optimal marching velocity for various loads in terms of metabolic cost, no studies have biomechanically assessed these limits. Investigating the effects of operationally relevant loads and marching velocities around the GTP on the coordination of gait during preferred-stride (running) and forced-march (walking) may reveal any suboptimal and potentially injurious gait patterns ergo identifying any adverse load-velocity combinations. These findings will have merit for informing load carriage training strategies in the military, especially in basic training and indoctrination courses where entry level recruits and candidates are abruptly exposed to some of the most arduous and intense training in the military.

1.3 Definition of the Problem

Load carriage is a crucial part of training and standard operations for military personnel. The modern-day warfighter must be able to carry their personal protective equipment (PPE), weapon systems and ammunition, and enough provisions to effectively carry out and accomplish missions in increasingly unpredictable and hostile environments. While commanders have years of experience to rely on and various strategies for determining what and how much gear/provisions are necessary, the unknown conditions of modern conflicts make these decisions difficult. Now that female applicants are being afforded the opportunity to fulfill combat roles, more considerations for how to best distribute the load amongst the troops may be necessary. Loads can become quite cumbersome, especially for extended training excursions or combat operations. Heavy backpack loads have been shown to significantly alter biomechanical aspects of posture and gait. These alterations increase stress placed on the musculoskeletal system and may consequently increases the risk for musculoskeletal injuries. In addition, the velocity at which the march is conducted has also been shown to significantly impact biomechanical aspects of gait. Forced marching raises concern as warfighters may be pushed to adapt less than optimal, and potentially injurious, gait patterns if sustained for prolonged periods of time. Any musculoskeletal injuries that result from load carriage may impact muscular force generation and force sustainment, hindering the warfighters ability to carry out essential squad tactics such as reaction and movement to contact, seeking cover, providing suppressive and cover fire, and executing fire and maneuver. Musculoskeletal injuries sustained by even a single warfighter during operations can reduce the integrity and combat effectiveness of the entire unit.

For the first time in US military history, female applicants are permitted to enlist in combat centric roles. Little is known about the effects of load carriage on the performance and physical wellbeing of female warfighters. Numerus studies have investigated the effects of load carriage mass and marching velocity on biomechanics in male subjects. However, few studies have investigated the effects of load carriage on female gait biomechanics, none of which have specifically looked at velocities around the GTP. Moreover, very little is known about the coordination and control of load carriage at a forced-march. Investigating the effect of load and locomotion pattern on the various biomechanical aspects of gait and posture may provide insight into the mechanism of acute and chronic injuries believed to be associated with load carriage. In addition, assessing the stability/instability in spatial and temporal patterns of movement sequencing while forced-marching with load may reveal how coordination is affected.

1.4 Purpose of the Study

The primary purpose of this study was to investigate the main effects of load carriage mass on kinematic, kinetic, and spatiotemporal gait characteristics at marching velocities around the GTP in female subjects. As a secondary aim, coordination patterns will be examined for each load and locomotion condition to determine levels of variability/stability within the system.

1.5 Specific Aims and Hypotheses

Specific Aim 1: Examine the effects of load carriage magnitude on kinematic and spatiotemporal gait characteristics at specific gait events during walking, running, and forced-marching.

Hypothesis 1: Increases in load carriage magnitude will decrease stride length, flight time, stance time, and increase stride width, stride frequency, double support time, and lower extremity joint flexion (ankle dorsiflexion, knee flexion, hip flexion) at heel-strike, mid-stance and toe-off.

Specific Aim 2: Examine the effects of load carriage magnitude on kinetic gait characteristics during walking, running, and forced-marching.

Hypothesis 2: Increases in load carriage magnitude will increase lower extremity joint moment at heel-strike, mid-stance and toe-off.

Specific Aim 3: Examine the effects of load carriage magnitude on lower limb coordination variability during walking, running, and forced-marching.

Hypothesis 3: Increases in load carriage magnitude will decrease variability of lower limb coordination patterns.

Specific Aim 4: Compare the coordination variability of running and forced-marching at the same velocity during load carriage.

Hypothesis 4: Lower limb coordination variability will be greater during running compared to forced-marching.

1.6 Study Significance

Musculoskeletal injuries afflict all branches of the military and cost the Department of Defense (DoD) millions of dollars in medical, healthcare, and disability expenses and millions of days of light/limited or no duty annually.(Feuerstein, Berkowitz, & Peck, 1997; Hearn, Rhon, Goss, & Thelen, 2017; Kaufman, Brodine, & Shaffer, 2000; Nindl, Williams, Deuster, Butler, & Jones, 2013) Backpack load carriage has been shown to adversely alter biomechanics, placing additional stress on the musculoskeletal system and augmenting the warfighters risk of sustaining a musculoskeletal injury. With the technological advancements and growing complexity of modern warfare, warfighters are continuously burdened with more gear. These heavier loads will conceivably induce greater musculoskeletal stress, magnifying the warfighters risk for sustaining a traumatic or overuse musculoskeletal injury.

With female applicants now being considered for combat occupations, there is a demand for studies investigating the effects of combat load carriage on the biomechanics of females. To date, no study has attempted to assess the effects of load carriage at marching velocities around the GTP on gait biomechanics in female subjects. In addition, very little is known about the coordination and control of lower-extremity movement patterns during load carriage. Investigating the biomechanical effects of current military load carriage demands is necessary for informing standard operating procedures in military field manuals and for providing insight into the mechanism of injury for those injuries believed to be associated with load carriage.

2.0 Review of Literature

The review of literature will provide an overview of previous research that relates to the current study. This section will further explore the epidemiology of military training related injuries, to include load carriage injuries. Next, the biomechanical and physiological aspects of load carriage will be covered followed by a discussion on gait transition and coordination as it relates to load carriage. Lastly. gender differences observed in various aspects of load carriage will be detailed.

2.1 Military Training Related Injuries

2.1.1 Injury Epidemiology

Non-fatal injuries represent one of the biggest threats to combat readiness and have been described as an under-recognized health problem in the military.(Hauret et al., 2010; B. Jones, Hansen, Kaufman, & Brodine, 1996; Bruce H Jones, Amoroso, Canham, Weyandt, & Schmitt, 1999) However, the DoD heavily prioritizes their efforts on fatal injuries, specifically motor vehicle crashes. Jones et al. conducted a broad, descriptive epidemiological study on injuries in the U.S. military.(Hauret et al., 2010) Data were collected from the Armed Forces Health Surveillance Center for the years 2000-2006. The relative magnitude of all acute/traumatic injuries and several health conditions were compared. Injury was defined as “traumatic wounds or other conditions of the body caused by external force or exposure (i.e., transfer of kinetic energy, heat, or cold), and microtraumatic physiologic harm resulting in loss of capacity due to a continued or repeated neuromuscular stress or strain".(Hauret et al., 2010) The definition included generally accepted ICDM-9-CM codes. Injuries encompassed traumatic injuries, overuse injuries (i.e. tendinopathy, bursitis, lumbago, sciatica, and thoracic/lumbosacral neuritis or radiculitis), stress fractures, sprains, strains, ruptures, dislocations, and other joint derangements (i.e. intervertebral disc disorders, meniscus tears, and joint instability). Injuries accounted for over 1.95 million medical encounters, more than 2.5 times the next leading category, mental disorders.(Hauret et al., 2010) Injury rates were reported for the DoD and calculated using person years in the denomination. The DoD injury rate was over 1600 injury visits per 1000 service members per year.(Hauret et al., 2010) Lower extremity overuse injuries were the most common type of injury in the DoD with rates nearing 900 injury visits per 1000 service members per year.(Hauret et al., 2010) Injuries reported were categorized according to the diagnosis and anatomical location. Of all musculoskeletal injuries examined in outpatient visits, 17.8% were of the lumbar spine, 28.2% were of the knee and lower leg, and 16.9% were of the ankle and foot.(Hauret et al., 2010) Overuse injuries were the most common injury type at 87.7% of all musculoskeletal injuries examined in outpatient visits.(Hauret et al., 2010) Among all musculoskeletal injuries that required hospitalization, 30.0% were of the lumbar spine, 28.0% were of the knee and lower leg, and 5.6% were of the ankle and foot.(Hauret et al., 2010) Overuse injuries accounted for 29.4% of all musculoskeletal injuries that required hospitalization.(Hauret et al., 2010) In comparison to the roughly 1,000,000 service members who suffered non-fatal, non-battle injures, there have only been a few hundred motor vehicle crashes and a few dozen aviation crashes that resulted in fatalities per branch each year.(Hauret et al., 2010) These data clearly indicate that non-fatal injuries are by far the biggest health problem of the military.(Hauret et al., 2010)

In another similar study, Hauret et al. focused more closely on the incidence of traumatic and overuse musculoskeletal injuries reported across the DoD in 2006. They collected data from military medical surveillance systems and evaluated all injury-related musculoskeletal conditions. Similar to Jones et al., musculoskeletal injuries were categorized according to their injury type and anatomical location. In total, 743,547 musculoskeletal injuries were reported in 2006.(Hauret et al., 2010) The injury rate was 628 injuries per 100 person-years.(Hauret et al., 2010) Among all musculoskeletal injuries observed in outpatient visits and hospitalizations, the lower extremities and lumbar spine were the most common injury sites at 39.0% and 19.5%, respectively.(Hauret et al., 2010) Further segmenting lower extremity injuries, the pelvis, hip and thigh comprised only 3.7% of all musculoskeletal injuries, the knee and shank comprised 22.4%, and the ankle and foot comprised 13.0%.(Hauret et al., 2010) Of all injuries reported, 82.3% were classified as overuse injuries.(Hauret et al., 2010)

Feuerstein et al. conducted a study on the prevalence of musculoskeletal-related disability in U.S. Army personnel focusing on gender differences and differences among Military Occupational Specialties (MOS).(Feuerstein et al., 1997) The study used the U.S. Army Physical Disability Agency's database and consisted of 41,750 case records spanning the years 1990 to 1994. Musculoskeletal disabilities were defined as "any physical disability case with a diagnosis of the following: skeletal/joint inflammatory disease; skeletal/joint impairments or limitation of motion; muscle injuries; or peripheral nerve neuritis or neuralgia in the hands, wrists, forearms, shoulders, neck, back, trunk, legs, knees, ankles, or feet".(Feuerstein et al., 1997) The relative frequency of various diagnoses and diagnostic frequencies for MOS and MOS by gender were calculated. To identify MOSs in which women experienced the greatest risk for developing a musculoskeletal disability, disability rates for women in specific MOSs were compared to the average disability rate for women in the Army. Additionally, female gender was identified as a risk factor if relative risk ratios were greater than 1.5. Results showed that women experienced higher rates of all physical disability and musculoskeletal-related disability compared to men.(Feuerstein et al., 1997) Musculoskeletal-related disabilities accounted for more than half of all disability cases diagnosed.(Feuerstein et al., 1997) The five most prevalent diagnoses reported were musculoskeletal limitation of motion, degenerative arthritis, lumbosacral strain, knee impairment and intervertebral disc syndrome.(Feuerstein et al., 1997) The top five MOSs with the highest disability incidence rates were Infantryman, Heavy Construction Equipment Operator, Man Portable Air Defense System Crew-member, Tube-Launched Optically-Guided Wire Missile Infantryman, and Mortar Crewman.(Feuerstein et al., 1997) Lastly, the top five MOSs where women displayed significantly higher disability incidence rates compared to men were Multi-Channel Transmission Operator, Single-Channel Radio Operator, Wheeled-Vehicle Mechanic, Signal Intelligence Analyst, and Construction Equipment Repair.(Feuerstein et al., 1997) These findings support previous research identifying a high prevalence of musculoskeletal injuries in the military.(B. Jones et al., 1996; B. Jones & Knapik, 1999) In addition, similar injury trends were reported.(B. Jones et al., 1996; B. Jones & Knapik, 1999) Furthermore, the higher overall disability rates in women and jobs where women have been identified as being at a greater risk for sustaining an injury indicate the need to research the differential factors between genders that may be contributing to these observed differences.

Ruscio et al. put the magnitude into a more relevant perspective, emphasizing the burden of musculoskeletal injuries on the DoD in terms of the number of limited duty days. In 2005, traumatic and overuse musculoskeletal injuries accounted for 24,918,244 days of limited duty.(Ruscio et al., 2006) The top five injures categorized by body region were lower extremity overuse, lower extremity fractures, upper extremity fractures, torso overuse, and lower extremity sprains, strains and ruptures. Lower extremity overuse injuries accounted for the greatest number of days of limited duty with an estimated 3,803,512 days of limited duty, nearly 15.3% of the total days of limited duty for the DoD.(Ruscio et al., 2006) From an absolute frequency perspective, the majority of injuries reported are non-fatal and are treated in outpatient settings. Although understanding the sheer magnitude of non-fatal injuries, associated health consequences, the amount of resources allocated to treat injuries, and time loss due to injury are all critical steps in determining the burden of non-fatal injuries on the DoD, it is equally as important to explore the leading causes of these non-fatal injuries. A clear understanding of the modifiable risk factors and causes of non-fatal injury are essential for the development and evaluation of injury prevention strategies. To attain leading injury causes, medical data was matched with each branches' mishap/safety reporting database. The match percentage for inpatient medical data was much higher than that of outpatient medical data, likely since inpatient records were more complete. The top five causes of hospitalizations for the leading types of injuries were falls, sports and physical training, guns and explosives, non-military vehicle related, and twists/turns/run/slips without a fall. Sports and physical training were the leading causes of lower extremity sprains, strains, and dislocations followed by falls and twists/turns/run/slips without a fall. Mitigation strategies focusing on sports and physical training, falls, and twists/turns/run/slips without a fall could make significant reductions in lower extremity sprains, strains, and dislocations.

2.1.2 Load Carriage and Injury

Studies specifically examining load carriage related musculoskeletal injuries fall into one of two categories; (1) studies documenting the incidence of load carriage injuries, and (2) studies investigating a single type of load carriage injury. As the load carried by the warfighter has increased, so has the incidence of musculoskeletal injuries. Studies reporting on the incidence of load carriage related injuries typically account for a single load carriage excursion. The collective findings from these studies report a wide injury incidence range of 24% to 90%.(Coopers, 1981; Dalén et al., 1978; J. Knapik et al., 1992) Distance traveled and loads varied between studies, which could explain the wide incidence range. Despite the discordant findings on injury incidence, similar injuries were reported across studies and most commonly involved the low back or lower extremities(Coopers, 1981; Dalén et al., 1978; J. Knapik et al., 1992), undoubtedly since these sites bear the brunt of the force. Other injuries, including superficial tissue damage from shoulder and hip straps and blisters on the plantar surface of the feet have been reported.(J. Knapik et al., 1996; J. Knapik et al., 2010; J. J. Knapik et al., 2004; Robin Marc Orr et al., 2014) While the latter may seem insignificant, they are still a cause for concern as more serious cases may alter biomechanics and potentially warrant limited duty or hospitalization. Medical problems commonly associated with load carriage include foot blisters, metatarsalgia, stress fractures, knee pain, and low back injuries.(J. Knapik et al., 1996)

Foot blisters are the most frequent injury sustained during a bout of load carriage. Again, although this injury may seem insignificant, blisters can develop into more serious medical conditions such as cellulitis or sepsis and, depending on the severity, they may trigger the warfighter to adapt an antalgic gait pattern. Any sustained gait abnormalities will likely alter muscle activation patterns and may place an atypical load on musculoskeletal structures not conditioned to withstand the additional load. More serious cases of foot blisters could contribute to musculoskeletal injuries reported during load carriage. A survey of U.S. Soldiers deployed to Iraq from the year 2003 to 2004 reported 33% of solider experienced foot blisters and 11% sought medical care.(Brennan, Jackson, Olsen, & Wilson, 2012) Frictional forces appear to be the mechanism of injury for foot blisters. As the warfighter marches, external forces as a result of locomotion attempt to move the sock and boot across the skin of the foot.(J. J. Knapik & Reynolds, 2016) Frictional forces will oppose this movement and with repeated force to the same location, skin temperature will increase forming “hot spots” and a blister may eventually develop.(J. J. Knapik & Reynolds, 2016) Studies show that as load carriage mass increases the likelihood of blister formation also increases.(J. Knapik et al., 1993; Reynolds, Kaszuba, Mello, & Patton, 1990) Other risk factors include magnitude of frictional force, number of shear cycles, skin characteristics, ethnicity, foot type, and condition of footwear.(J. J. Knapik & Reynolds, 2016)

Metatarsalgia is a term used to describe a nonspecific overuse injury of the foot. Symptoms include localized pain or tenderness on the sole of the foot, especially under the second, third, and fourth metatarsal heads.(J. J. Knapik & Reynolds, 2016) Rapid changes in the intensity of weight bearing activity can cause foot strain injuries that are often associated with metatarsalgia.(J. J. Knapik & Reynolds, 2016) Sutton et al. observed a 7-month Army Airborne Ranger physical training program that included load carriage and reported on athletic injuries and performance limiting conditions.(Sutton, 1976) The incidence of metatarsalgia was 20%.(Sutton, 1976) While the exact mechanism of injury is unknown, it has been postulated that the excess stress from the additional load as the foot rotates anteroposteriorly around the distal ends of the metatarsal bones during marching can induce inflammation and trauma to the metatarsal heads.(J. J. Knapik & Reynolds, 2016) For this reason, load carriage mass is believed to be as a risk factor for metatarsalgia.(J. J. Knapik & Reynolds, 2016)

As discussed in the previous section, lower extremity stress fractures are common in military training. A stress fracture is an incomplete bone rupture often associated with repetitive overloading and eventual mechanical deformation of bones. In a retrospective cohort study, Knapik et al. investigated stress fracture risk in a large sample of Army basic training recruits over an 11 year period.(J. Knapik et al., 2012) The results showed incidence rates of 1.9% and 8.0% for men and women respectively.(J. Knapik et al., 2012) An odds ratio revealed women were nearly four and a half times more likely to sustain a stress fracture during basic training.(J. Knapik et al., 2012) In the military, stress fractures were previously referred to as “march fractures” since the majority of cases presented following a ruck march.(Hamilton & Finklestein, 1944) Stress fractures are classified as overuse injuries and generally occur in those individuals with otherwise healthy bones during repetitive, unaccustomed physical activity. The current understanding is that stress fractures occur when the mechanical stress and osteoclastic processes, or bone resorption, outweigh the osteoblastic processes, or bone remodeling.(J. J. Knapik & Reynolds, 2016) These conditions may arise during a prolonged load carriage task, especially in those individuals not accustomed to the additional load. Several risk factors for the development of stress fractures have been identified and are well documented within the literature. These risk factors include female gender(B. Jones & Cowan, 1993; B. H. Jones, Harris, Vinh, & Rubin, 1989; J. Knapik et al., 2012; Mattila, Niva, Kiuru, & Pihlajamäki, 2007), white ethnicity(J. Knapik et al., 2012; Mattila et al., 2007; Shaffer, Brodine, Almeida, Williams, & Ronaghy, 1999), older age(J. Knapik et al., 2012; Mattila et al., 2007; Shaffer et al., 1999), taller body stature(B. Jones & Cowan, 1993; J. Knapik et al., 2012), lower aerobic fitness(B. Jones & Cowan, 1993; Shaffer et al., 1999), prior physical inactivity(B. Jones & Cowan, 1993; Shaffer et al., 1999), and higher amounts of current physical training.(B. Jones & Knapik, 1999; Popovich, Gardner, Potter, Knapik, & Jones, 2000)

Knee pain is another condition often reported in association with load carriage. Knee pain is a broad term that could originate from several pathological conditions including patellofemoral pain syndrome, patellar tendonitis, bursitis, and ligamentous strain. In a study on injuries recorded during strenuous 20km road march with a 46kg load, Knapik et al. reported a 1.2% incidence of post-march knee pain.(J. Knapik et al., 1992) Reynolds et al. collected injury data on light infantry soldiers over a 161km road march with an average load carriage mass of 47 ± 5kg. The 5-day load carriage training exercise resulted in a 2.5% incidence of post-march knee pain. In another study that reported on load carriage injuries of Australian Army Soldiers, 11% of load carriage related injuries involved the knee.(Robin M. Orr, Johnston, Coyle, & Pope, 2015) Knee pain that occurs following a bout of load carriage is likely a result of the repetitive mechanical exchange of energy. Increases in load carriage mass (from 20 to 50% of body weight) are associated with increases in lower extremity joint angles and a decrease in the body's center of mass.(Attwells et al., 2006; E. Harman et al., 2000; Kinoshita, 1985) These biomechanical adaptations are believed to serve a protective function, increasing joint angles to absorb the impact forces and lowering center of mass to increase stability. However, a number of studies have found that at loads of 50% of body weight or higher, there is little further increase in knee joint angle.(Attwells et al., 2006; E. Harman et al., 2000; Kinoshita, 1985) In addition, as load carriage mass increases, knee joint moments are seen to increase.(E. Harman et al., 2000; Wang, Frame, Ozimek, Leib, & Dugan, 2013) These findings suggest that with loads nearing 50% of body weight, the knee joint will reach a ceiling for range of motion and presumably impact absorbing effects, which may subject the joint larger moments. The combination of increased range of motion and greater forces at the knee joint during heavy load carriage may be associated with knee pain. Identified risk factors for knee pain include greater patellar mobility, lower quadriceps flexibility, faster vastus medialis reflex time, lower vertical jump performance, greater hip adduction, lower isometric strength in hip adduction, knee extension and knee flexion, greater navicular drop, and a smaller knee flexion angle.(Boling, 2009;

Kujala, Kvist, Osterman, Friberg, & Aalto, 1986; Witvrouw, Lysens, Bellemans, Cambier, & Vanderstraeten, 2000)

Back pain and low back injuries are frequently reported following load carriage tasks. The origin of back pain is difficult to define, as there are several pathological conditions and variety of structures for witch the pathology may arise. In the same study on a strenuous 20km road march, Knapik et al. observed that 23% injuries sustained during the road march involved the back.(J. Knapik et al., 1992) Additionally, 50% of the soldiers that were unable to complete the march reported experiencing problems with their back.(J. Knapik et al., 1992) Orr et al. found back injuries to be the most common injury sustained during load carriage in Australian Army Soldiers, accounting for 23% of all injuries.(Robin M. Orr et al., 2015) As the warfighter marches, cyclical stresses are place on the vertebrae, intervertebral disk, and the muscles and ligaments stabilizing the spine. As load increases, so do the compressive forces, shear forces, and torques about the spine. The magnitude of these forces are dependent on load carriage mass and are believed to be associated with back pain and back injuries. Load carriage mass and back injuries appear to share a dose response relationship. The greater the load, the more back injuries observed.

2.2 Biomechanical and Physiological Aspects of Load Carriage

Load carriage and marching velocity are among two of several variables effecting the biomechanical and physiological aspects of gait. The effects of each variable on gait have been studied in some depth. This section will cover the specific biomechanical and physiological effects of load carriage mass, marching velocity, as well as the interaction effects of load-velocity. The load carriage mass and marching velocity subsections will begin with the biomechanical aspects and course into the physiological aspects. The interplay between the biomechanics and physiology is often reciprocal, meaning that changes in one will impact the other.

2.2.1 Load Carriage Mass

The most prominent kinematic effect of backpack load carriage is an increased forward lean of the body.(Martin & Nelson, 1986) The effect of various load carriage methods and weight distribution on biomechanics has been studied thoroughly, and results have shown that load positioning within a pack will impact the degree of forward body lean. The lower that the load is positioned in the pack, the more forward rotation about the hips or ankles that is required to bring the pack center of mass (CoM) over the feet.(Bloom & Woodhull-Mcneal, 1987; Johnson, Pelot, Doan, & Stevenson, 2000) The mass of the load also effects inclination of the body.(Attwells et al., 2006; Martin & Nelson, 1986) With heavier loads, increases in forward lean of the body are observed.(Attwells et al., 2006; Martin & Nelson, 1986) In addition, the further the CoM of the load is away from the body, the greater the forward lean to offset the load.(Abe et al., 2008) Therefore, heavier loads positioned low within the backpack and away from the body will require a considerable forward lean, further shifting the body's CoM over the front half of the feet and possibly increasing the risk for foot injuries.(Abe et al., 2008; Attwells et al., 2006; Bloom & Woodhull-Mcneal, 1987; J. Knapik et al., 2010; Martin & Nelson, 1986) Bloom and Woodhull-McNeal conducted a study on the postural adjustments of standing with two different load carriage systems and found that higher load placements elicit less forward lean of the trunk at the expense of postural stability, especially for taller individuals.(Bloom & Woodhull-Mcneal, 1987) This trend was also observed in dynamic load carriage tasks.(Bobet & Norman, 1984) Furthermore, higher load placements were also shown to increase the activation of erector spinae and upper trapezius muscles and greater muscular force was required to correct for linear and angular acceleration of the backpack.(Bobet & Norman, 1984) The data suggests that mid to low back load placements may be safer for marches over uneven terrain where stability is paramount. A stumble or misplaced placed step will generate linear and angular acceleration of the backpack and trunk.(Bobet & Norman, 1984) Recovering from a stumble and maintaining postural stability will require relatively lower muscular forces with a lower load placement.(Bobet & Norman, 1984) Higher load placements may be beneficial over even terrain as they entail a lower energy cost and less forward lean of the trunk.(Bloom & Woodhull-Mcneal, 1987; Johnson et al., 2000; Obusek, Harman, Frykman, Palmer, & Bills, 1997; Stuempfle et al., 2004)

Significant alterations to certain spatiotemporal gait parameters are seen during load carriage tasks. In general, as load increases at a fixed pace, decreases in stride length and swing duration can be seen with increases in stride frequency and double-support, providing greater stability.(Attwells et al., 2006; Kinoshita, 1985; Martin & Nelson, 1986) Several studies have reported no change in stance duration with increases in load, however, swing time was seen to decrease significantly with increases in load up to 50% of bodyweight.(Ghori & Luckwill, 1985; Kinoshita, 1985; Martin & Nelson, 1986) The decrease in swing time and maintenance of stance time accounts for the increase of double-support time. Pelvic rotation is seen to decrease while hip excursion increases to compensate. However, the increase in hip excursion is not enough to fully counteract the decrease in pelvic rotation and consequently stride length shortens and stride frequency increases to keep pace.(M. LaFiandra, Wagenaar, Holt, & Obusek, 2003) Despite these findings, there is contradictory evidence as to whether most individual shorten their stride length.(E. Harman et al., 1992) Nonetheless, a preponderance of the evidence suggests that stride length will decrease or remain constant with increases in external load.(Attwells et al., 2006;

Kinoshita, 1985; Martin & Nelson, 1986) Kinoshita et al. investigated the effects of load carriage on select biomechanics of walking gait and observed increased knee flexion with heavier loads. This was proposed to be a shock absorbing mechanism to reduce impact forces and allow for a smoother transfer of weight to the supporting surface(Kinoshita, 1985) However, the subjects were not regularly engaged in load carriage tasks. Tilbury-Davis and Hooper found no such phenomena with increasing loads.(Tilbury-Davis & Hooper, 1999) Their subjects were active duty military personnel who were regularly engaged in load carriage tasks. The difference in findings between the studies may be attributed to the dissimilarity in task familiarization of the participants. Those with load carriage experience likely have stronger lower extremity musculature and more developed patterns of movement as a result of training. The experienced warfighter will likely exhibit different, and possibly more efficient, biomechanical characteristics in response to load. Nonetheless, the alterations seen in gait with increases in load are considered positive adaptations. However, the resultant increase in lower extremity joint forces and torques are an unavoidable consequence and may increase the risk of sustaining a musculoskeletal injury.

Regarding the kinetics of load carriage, increases in ground reaction forces (GRF), breaking forces (BF), propulsive forces (PF), and lateral forces (LF) are observed with increases in load.(E. Harman et al., 1992; Kinoshita, 1985; Tilbury-Davis & Hooper, 1999) In the same study conducted by Tilbury-Davis and Hooper, investigators also examined the effects of load carriage on lower extremity kinetics.(Tilbury-Davis & Hooper, 1999) Their findings indicated that increases in load carriage mass increased ground reaction forces in proportion to the subjects' total mass (body mass plus load carriage mass).(Tilbury-Davis & Hooper, 1999) These results are in conflict with previous research which found no such proportional relation between ground reaction forces and load carriage mass.(E. Harman et al., 1992) Such differences may again be explained by the inclusion of trained subjects in the Tilbury-Davis et al. study. Additionally, the difference in marching velocity between studies may also contribute to the difference in their results. Peak forces were seen to increase with the lightest load of 20kg and forces necessary for balance increased significantly for all load carriage conditions.(Tilbury-Davis & Hooper, 1999) Interestingly, lower extremity impact forces from the unloaded condition to the 20kg condition were observed to decrease and impact forces from the unloaded condition to the 40kg condition were not significantly different.(Tilbury-Davis & Hooper, 1999) This attenuation of lower extremity loading and unloading rates when bearing heavier loads may indicate the presence of a protective mechanism. In addition to the forces experienced by the lower extremities, mechanical loading of the spine has also been shown to increase with increases in load.(Goh et al., 1998) Goh et al. examined the effects of backpack load carriage on peak forces in the lumbosacral spine, finding that peak forces increased disproportionately with increasing load.(Goh et al., 1998) When compared to level walking without load, walking with backpack loads of 15 and 30% of body weight increased peak lumbosacral forces by 26.7 and 64%, respectively.(Goh et al., 1998) Lumbosacral forces were predominately compressive forces with a lower magnitude shear force components.(Goh et al., 1998) The large increases in lumbosacral forces observed with a given increase in load carriage mass is a cause for concern in military populations, as load carriage mass often exceeds 30% of the individuals body weight.

Renbourn, former Major in the Royal Army Medical Corps, stated “the load carried by the soldier … will probably always be a compromise between what is physiologically sound and what is operationally essential”.(Renbourn, 1954) Numerous studies have been conducted on the physiological aspects of load carriage. While not the primary aspect of interest for the current study, the biomechanics of load carriage will impact physiology and vice versa. Studies have found the distribution of load across the body to have a pronounced effect on energy cost. In general, the closer the load is to the body's center of mass (COM), the lower the energy cost. Studies investigating the energy cost of various load carriage methods have found that head-supported load carriage is the most efficient for transporting a load.(J. J. Knapik et al., 2004; Willems, Heglund, Cavagna, & Penta, 1995) For obvious reasons, this method is not practical for the warfighter. A more sensible method with markedly low energy cost is the double-pack.(J. Knapik et al., 1996) Double-packs evenly distribute the load from the front to the back of the carrier and produce fewer deviations from normal gait than a backpack.(E. Harman et al., 1992) Under some circumstances the double-pack may be useful (medics/non-combatants, non-combat operations), however, in most operational scenarios it will be a functional hindrance to the warfighter. Having a portion of the load hanging from the front of the carrier will likely limit their field of view and restrict their movement. As is apparent, navigating over uneven terrain with a limited field of view will very likely increase injury risk. Moreover, heavier double-pack loads may obstruct ventilation and consequently have a negative effect the warfighters ability to endure physical tasks.

Backpack load carrying is widely accepted as the most appropriate manner for increasing load carriage capacity of the warfighter. While this load carriage method may entail greater energy cost than the double-pack(J. Knapik et al., 1996), the warfighter is more mobile and capable of executing mission essential tasks. Some research has shown that the positioning of the load in the backpack itself will have an effect on energy cost.(Obusek et al., 1997; Stuempfle et al., 2004) Loads positioned low in the backpack and away from the body were associated with high metabolic costs. One study found no statistically significant difference in oxygen consumption between backpack load placements (high, middle, and low position) and a load carriage vest that distributed the load evenly from front to back of the carrier, much like a double-pack.(Johnson et al., 2000)

However, the authors ascribe the results of their study to the short duration of the load carriage protocol, suggesting that a longer bout of load carriage may present differences in oxygen consumption between the load placement conditions. The consensus within the literature is that the energy cost of load carriage increases in a systematic manner with increases in body weight, load carriage mass and/or velocity.(Keren et al., 1981; Soule et al., 1978) Interestingly, Abe et al. observed an energy saving phenomenon known as “free-ride” during backpack load carriage at slower speeds.(Abe et al., 2004) Free-ride was previously observed in studies of head-supported load carriage and researchers found that metabolic cost did not increase with loads below 20% of the individuals’ body weight.(Willems et al., 1995) Abe et al. witnessed a similar phenomenon that was most evident when subjects were carrying a backpack load of 9 kg at speeds lower that 80 m/min.(Abe et al., 2004) However, it is not likely that this phenomenon occurs in training or combat operations, as warfighters are continually outfitted with loads much heavier than 9kg and required to march at speeds faster than 80 m/min.

2.2.2 Marching Velocity

The most prominent adaptations to increases in marching velocity are increases in stride length and stride frequency. With increases in velocity, increases in lower extremity joint ROM are observed. The body’s CoM becomes lower as a result of increased hip flexion, knee flexion, and ankle dorsiflexion. The percent of time in stance phase progressively decreases from walking to running. Once velocity has risen to a critical value where stride length and stride frequency reach a ceiling, further increases in velocity will induce a transition to running, as the double support phase diminishes, and a flight phase emerges.

Murray et al conducted a study examining the biomechanics of walking at a self-selected pace (1.51±0.20 m/sec) and a fast-pace (2.18±0.25 m/sec) in 30 male subjects, with mean values of stride frequency for self-selected pace and fast-pace walking of 113 and 138 steps/min, respectively.(Murray, 1967) The mean values for stride length in both the self-selected pace and fast-pace conditions were 1.56±0.13m and 1.86±0.16m, respectively.(Murray, 1967) Increases in velocity resulted in decreases in duration of stance, swing and double-support phases.(Murray, 1967) However, and contrary to the effects of increasing load, the duration of stance phase was seen to decrease 3.5 times as rapidly as swing phase with increases in velocity.(Murray, 1967) The mean durations of stance phase for self-selected pace and fast-pace walking were 0.65±0.07 sec. and 0.49±0.05 sec., respectively.(Murray, 1967) The mean durations of swing phase for self-selected pace and fast-pace were 0.41±0.04 sec. and 0.38±0.03 sec., respectively. (Murray, 1967) As a result of this more rapid decrease in stance phase, the percentage of stride in stance decreased from 61% to 57% but the percentage of stride in swing increased from 39% to 43% from self-selected pace to fast pace.(Murray, 1967) Swing phase duration seems to be a strong determinate in achieving faster walking speeds, as the swinging limb must travel greater distances in less time to successfully increase stride length and stride frequency.(Murray, 1967) The mean durations of double-support phase for self-selected pace and fast-pace were 0.12±0.03 sec. and 0.06±0.03 sec., respectively.(Murray, 1967)

2.2.3 Interaction Effects of Load Carriage Mass and Marching Velocity

While studies investigating the effect of a single perturbation on gait provide significant insight on the implications of each, it's important to consider that these variables almost always coincide and often elicit different biomechanical and physiological effects when interacting on the body simultaneously. As discussed in the previous section, several studies have examined the biomechanical effects of load carriage mass while controlling for marching velocity and vice versa. Only a few studies have investigated how load and speed interact in their effects.

After conducting two separate studies, one on the biomechanical effects of load carriage mass(E. Harman et al., 2000) and the other on marching velocity(E. A. Harman et al., 2000), Harmen et al. set out to explore the interaction effects of load and velocity in a subsequent study.(E. A. Harman et al., 2001) They recruited 16 male subjects, a number of which were active duty soldiers in the U.S. Army.(E. A. Harman et al., 2001) The study utilized force plates, accelerometers, a motion capture system, and electromyography to capture data on ground reaction forces, backpack peak accelerations, joint ranges of motion, joint torques, joint forces, spatiotemporal gait parameters, peak and average muscle activity, and timing of muscular activation.(E. A. Harman et al., 2001) Subjects were outfitted with 4 backpack loads (6, 20, 33, and 47kg) and required to walk at 3 different velocities (1.1, 1.3, and 1.5m/s) for a total of 12 possible load-speed combinations.(E. A. Harman et al., 2001) Backpack loads were split up into four separate testing sessions and each session comprised 3 trials at each velocity for a total of 9 trials.(E. A. Harman et al., 2001) For each trial, subjects walked 15m across a force platform in the motion capture field of view. The biomechanical effects of load were consistent with previous research.(E. A. Harman et al., 2001) The biomechanical effects of velocity were also in accordance with previous research.(E. A. Harman et al., 2001) There were few statistical interaction effects reported, indicating that increases in load had the same effects on gait over all three velocities and increases in velocity have the same effects on gait over all four backpack loads.(E. A. Harman et al., 2001) Many of the effects were in the same direction meaning that increases in velocity and load resulted in an increase of the variable.(E. A. Harman et al., 2001) However, the effects of load and velocity were in opposition for a few variables.(E. A. Harman et al., 2001) For those variables, the effects of speed and load cancelled each other out. Those variables that exhibited a statistical interaction were stride frequency, stride time, and downward impulses.(E. A. Harman et al., 2001) At 1.17 and 1.33 m/s velocities, stride frequency increased and stride time decreased when the load increased from 33kg to 47kg, but no such adaptations occurred at 1.5m/s.(E. A. Harman et al., 2001) Impulse increased with heavier loads but decreased with increasing speed.(E. A. Harman et al., 2001) The increase in impulse was directly related to pack weight except for the 1.33m/s velocity, where increases in impulse were less than proportional to the increase in pack weight from 33 to 47kg.(E. A. Harman et al., 2001) Stride time and stride frequency are mathematical inverses of each other and impulse is sensitive to changes in stride time. All three variables are related. At velocities of 1.17 and 1.33m/s, subjects were able to adapt to the heaviest load by decreasing stride length and increasing stride frequency, thus decreasing stride time.(E. A. Harman et al., 2001) These are considered favorable adaptations that serve to promote a stable base of support and attenuate the impulse about the lower extremity joints. However, these impulse-reducing adaptations were not seen at 1.5 m/s.(E. A. Harman et al., 2001) With the heaviest load at the highest velocity, subjects experienced significantly higher downward impulses.(E. A. Harman et al., 2001) Increase of either variable, load or velocity, resulted in increased musculoskeletal stress.(E. A. Harman et al., 2001) In addition to the higher downward impulses, bone-on-bone forces, ground reaction forces, and muscle electrical activity all increased with increases in load or velocity.(E. A. Harman et al., 2001) During a military road march, a sustained combination of these factors may augment the warfighters risk of injury.

Liew et al. followed a similar methodology, however, they tested male and female subjects using lighter loads (0%, 10%, and 20% of body weight) at running velocities (3, 4, 5 m/s).(Liew, Morris, & Netto, 2016) Using force platforms and a motion analysis system, they collected data for ground reaction forces and lower limb trajectories.(Liew et al., 2016) Relationships between load and joint power and angle vectors at each velocity were analyzed.(Liew et al., 2016) In agreement with previous studies, results showed that load was positively correlated to joint power in the second half of stance, ankle angles during mid-stance (4.0 and 5.0 m/s), knee angles at mid-stance (at 5.0 m/s), and hip angles during toe-off (all velocities).(Liew et al., 2016) At the faster running velocities, increases in load appeared to alter power contribution of the lower extremity joints in a distal-to-proximal manner from mid-stance to toe off.(Liew et al., 2016) Kinematic changes were multiplanar, as evidenced by the influence of load on both sagittal and frontal plane lower extremity joint angles.(Liew et al., 2016) Mechanically, running predominantly occurs in the sagittal plane, however, forces applied to the body act in three dimensions and therefore must be counteracted in three dimensions. The changes in joint power and kinematics observed with load carriage during running will also increase musculoskeletal stress. Although the prescribed loads were lower than combat loads used in the military and the velocities were much higher than that of a road march, the results of this study indicate that the body responds to increases in load and velocity in a similar manner for both walking and running. Additionally, there may be an upper limit for marching velocity while carrying a certain load, at which point any further increase in velocity will result in a significantly greater increase in musculoskeletal loads and the energy cost of load carriage.

2.3 Load Carriage Gait Transition and Coordination

The gait patterns of walking and running are easily characterized by their spatial and temporal symmetry. Both gait patterns are well understood and are commonly defined either dynamically through their distinct fluctuations of potential energy and kinetic energy or spatiotemporally through the presence or absence of a flight phase.(Segers, 2006) While concepts around each gait pattern are well studied and adequately understood, the transition point between the two gait patterns and how humans realize the transition from one gait to the other is no quite as clear.(Segers, 2006) Moreover, even less is known about the effects of body borne loads on gait transition and lower limb coordination. Most of the literature suggests that the transition occurs in as little one stride(Segers, 2006), however, some research argues that there is an evident period of destabilization, or a non-equilibrium phase, prior to and post the steps around the transition from walking to running.(Diedrich & Warren, 1995) It has been suggested that each gait corresponds to a stable attractor and the non-equilibrium phase transition reflects a loss of stability and behaves as a bifurcation between the two attractors (walking and running).(F. J. Diedrich & W. Warren, 1998; Diedrich & Warren, 1995; Segers, 2006) Following this non-equilibrium phase transition, the system re-organizes itself into a more stable and coordinated pattern of movement. Farley and Ferris indicated that investigating the motor control mechanisms of the transition between gaits may help to improve our understanding of human locomotion.(Claire T. Farley & Daniel P. Ferris, 1998)

Reducing mechanical stresses and minimizing energy costs are two of the most common theories for the gait transition mechanism. The hypothesis of a ‘mechanical trigger’ suggests that the transition from one gait to another occurs to reduce mechanical stress and avoid overuse or injury.(F. J. Diedrich & W. Warren, 1998) Hreljac et al. proposed that the transition between gaits occurs to prevent overexertion of ankle dorsiflexors.(Hreljac, 1995) However, the activity of ankle dorsiflexors does not accurately predict the gait transition point (GTP), as the walk-to-run transition (WRT) will likely increase the exertion of the dorsiflexors.(F. J. Diedrich & W. Warren, 1998) The ‘energetic trigger’ theory suggests that gait transition occurs as a direct result of energy optimization.(F. J. Diedrich & W. Warren, 1998) Much like the previous theory, energy cost has not been shown to predict the GTP with sufficient accuracy, as findings within the literature vary. An alternative to these optimization criteria, with growing evidence, is the dynamical systems theory (DST).(F. J. Diedrich & W. Warren, 1998; Diedrich & Warren, 1995; Segers, 2006; Turvey, 1990) The DST views the resultant movement as a consequence of the boundary of constraints formed by the demands of the task, the capabilities of the organism, and the conditions of the environment.(F. J. Diedrich & W. Warren, 1998; Diedrich & Warren, 1995; Segers, 2006; Turvey, 1990) In contrast to previous theories of motor programs imposing organization on movement systems, the DST portrays human movement as a highly complex network of codependent subsystems that work concurrently to execute a desired movement pattern.(Turvey, 1990) Interaction between subsystems and exploration of the organismic, task, and environmental constraints for a goal-directed action drives the evolution of coordinated movement. DST considers stable coordinative movement patterns and the transitions between them a consequence of self-organization of the movement system.(F. J. Diedrich & W. Warren, 1998; Diedrich & Warren, 1995; Segers, 2006; Turvey, 1990) Principles of self-organization have long been applied to other physical and chemical sciences to explain the spontaneous emergence of an organized pattern of activity within a system.(Turvey, 1990) Therefore, much of the organization of a coordinative movement pattern may arise more locally rather than from the central nervous system. Several studies provide evidence for a dynamic interpretation the WRT.(F. J. Diedrich & W.

Warren, 1998; Diedrich & Warren, 1995; Taga, 1995a, 1995b; Thelen, Ulrich, & Society for Research in Child, 1991; Turvey, 1990; Turvey, Holt, Obusek, Salo, & Kugler, 1996) Furthermore, manipulating the task dynamics (i.e. increasing treadmill grade, adding ankle weights) has been shown shift the location of the stable attractor.(F. J. Diedrich & W. H. Warren, 1998) It is within reason to speculate that combat load carriage would also have an effect on preferred walking and running velocities and the transition point between gaits.

Without a load, the WRT is seen to occur spontaneously at approximately 2.0 m/s, although a walking gait can be maintained at higher velocities. However, the effect of load carriage on the GTP has not been formally studied. Keren et al. conducted a study on the energy costs of walking a running with load in 15 physically active males.(Keren et al., 1981) Investigators observed a critical point at which the energy cost of walking was relatively equal to the energy cost of running at the same velocity.(Keren et al., 1981) That is to say, the subject would expend more energy if they did not transition to running with further increases in velocity. Mercier et al. observed this same energy cost mechanism in later research.(Mercier et al., 1994) This effect was seen to be more pronounced during load carriage, especially if the load was a greater percentage of the subjects body weight.(Keren et al., 1981) Although gait transition velocity was not an outcome measure of direct interest for the study, investigators also collected data on the upper limits of walking velocity for both unloaded and loaded conditions.(Keren et al., 1981) Interestingly, they found that for the unloaded and loaded condition the upper limits for walking velocity were 2.26 and 2.16 m/s, respectively.(Keren et al., 1981) The results would suggest that controlling the rate of march, especially for a load carriage task, is essential for optimizing efficiency. What’s more, the findings indicate that load carriage may have an impact on the GTP. However, subjects were civilian males with no prior load carriage experience. Adequate load carriage training may alter energy optimization mechanisms and the GTP. Nevertheless, these findings merit further investigation into the effects of load carriage on the GTP.

LaFiandra et al. conducted a study on the influence of load carriage and walking speed on trunk coordination and stride parameters.(M. LaFiandra et al., 2003) Investigators hypothesized that load carriage would affect the intersegment coordination of pelvic and thoracic rotation.(M. LaFiandra et al., 2003) During unloaded walking, pelvic and thoracic rotation gradually transition from a more in-phase pattern to an out-of-phase pattern at velocities above 0.8 m/s.(M. LaFiandra et al., 2003) However, no such phenomenon was observed during load carriage.(M. LaFiandra et al., 2003) Counter-rotation of the pelvis and thorax during unloaded walking at higher velocities is believed to reduce the net angular momentum of the body.(M. LaFiandra et al., 2003) LaFiandra et al. speculate that counter-rotation as a means of reducing net angular velocity may not be necessary during load carriage.(M. LaFiandra et al., 2003) The addition of a backpack may increase the upper body's transverse plane moment of inertia, and as a result, less thoracic angular velocity may generate enough angular momentum to counterbalance the angular momentum of the lower body.(M. LaFiandra et al., 2003) However, counterbalancing is observed in the reciprocal motion of arm and leg swinging during load carriage.(M. LaFiandra et al., 2003) The authors suggest that increasing load carriage walking velocity above 1.61 m/s may require counter rotation between the pelvis and thorax.(M. LaFiandra et al., 2003) In a separate study conducted by Yen et al., investigators theorized that low back pain attributed to load carriage may be a result of decreased coordination variability producing mechanical stress.(Yen, Gutierrez, Ling, Magill, & McDonough, 2012) Contrary to their hypothesis, they found that load carriage increased coordination variability of trunk and thigh segments as well as the pelvis and the thorax in both sagittal and transverse planes.(Yen et al., 2012) It is evident that load carriage effects coordination patterns of the upper body. Changes in the coordination of pelvic and thoracic rotation seen during load carriage may, in some measure, account for the high incidence of back pain and low back injuries attributed to load carriage.

2.4 Sex Differences

When walking without load, men and women tend to employ similar gait kinematics and kinetics. However, with the addition of a backpack load, studies have shown that women walk with shorter stride length and greater stride frequency when compared to men.(Ling, Houston, Tsai, Chui, & Kirk, 2004; Martin & Nelson, 1986) As the load increases, women's stride length has been shown to significantly decrease(Ling et al., 2004; Martin & Nelson, 1986), while some studies have shown no significant stride length changes in men.(E. Harman et al., 1992) Additionally, women show a steeper linear increase in double support time with increasing load. Furthermore, women tend to have a greater forward trunk lean and hyperextended neck in order to shift the pack CoM over their feet.(Krupenevich, Rider, Domire, & DeVita, 2015; Ling et al., 2004; Martin & Nelson, 1986) However, in studies where loads were normalized to a percentage of body weight, men and women adopted similar gait adaptations.(Silder, Delp, & Besier, 2013) One study found no significant differences in lower extremity kinematics and kinetics between men and women when using a standardized load, suggesting that the absence of sex differences in their study may be a contributing factor to the increased injury incidence observed in female warfighters.(Krupenevich et al., 2015) However, the protocols for these studies were too short to elicit and account for any biomechanical adaptation as a result of fatigue. Sex differences may be more pronounced after a prolonged bout of load carriage. Regardless of the presence or absence of sex differences documented in the literature, response to heavy loads for extended durations will likely pose different, and potentially more adverse consequences for the female warfighter.

Several studies have been conducted on sex differences in injury rates, finding that female warfighters experience an overall higher injury incidence during recruit training and fleet force military training.(Piantanida, Knapik, Brannen, & O'Connor, 2000; Strowbridge, 2002) However, research on sex differences in load carriage injures among military personnel is scarce. In an effort to identify sex differences in load carriage injuries, Orr et al. conducted a retrospective cohort study to determine relative risks and patterns of load carriage related injuries in male and female Australian army soldiers.(R. M. Orr & Pope, 2016) Occupational health and safety incident data for active duty Army personnel were identified and extracted for the years 2009 and 2010.(R. M. Orr & Pope, 2016) The incident data were then paired with active duty Army population data for the same years.(R. M. Orr & Pope, 2016) Any incidents relating to injuries reported during load carriage events were extracted.(R. M. Orr & Pope, 2016) Load carriage events were defined as any activity where the soldier reported wearing webbing equipment, personal protective equipment or backpack, or where the specific activity at the time of the injury clearly indicated a load carriage activity.(R. M. Orr & Pope, 2016) Serious personal injuries were defined as injuries that required immediate treatment and all other injuries not requiring immediate treatment were classified as minor personal injuries.(R. M. Orr & Pope, 2016) Their findings indicated that overall levels of load carriage injury risk were not discernibly different between male and female soldiers.(R. M. Orr & Pope, 2016) The back was the leading site injury for both male and female soldiers followed by the ankle, knee and 'neck and shoulder' complex.(R. M. Orr & Pope, 2016) However, foot injuries reported among female soldiers occurred at more than twice the rate (IRR: 2.37) of their male counterparts.(R. M. Orr & Pope, 2016) In addition, the results show that female soldiers had twice the level of risk for sustaining a serious personal injury during a load carriage task.(R. M. Orr & Pope, 2016) Despite the similarities in overall load carriage injury rates, the authors noted that the study was conducted just before removal of gender restrictions for certain military occupations. With female soldiers now being permitted to assume combat centric roles, they are more likely to be exposed to heavier loads. It could be conjectured that female soldiers would sustain more severe injuries at a greater incidence when heavier loads are carried.

In addition to the sex differences in biomechanics and injury incidence during load carriage, there are a number of physiological and performance related differences. Female warfighters typically have a lower absolute maximal aerobic capacity than their male equivalents.(Allison et al., 2015) Therefore, to complete a given load carriage task, female soldiers would have to perform at a higher percentage of their maximal aerobic capacity when compared male soldiers.(Allison et al., 2015) Moreover, several studies investigating risk factors for military training related injuries have associated lower levels of aerobic fitness with an increased risk of injury.(B. Jones & Cowan, 1993; Shaffer et al., 1999) In drawing a connection between these studies, it is reasonable to speculate that female warfighters may be at a greater risk for sustaining an injury during a load carriage task due to the increased metabolic stress they endure. Harper et al. investigated various aspects related to load carriage performance in male and female soldiers to include maximal effort road march times, performance of upper and lower body tasks following a maximal effort road march, heart rate, ratings of perceived exertion, compatibility of current load carriage equipment, incidence of blisters, and pain, soreness and discomfort levels.(Harper, Knapik, & De Pontbriand, 1997) 19 male and 15 female soldiers completed the three phases of the study(Harper et al., 1997). Phase 1 was dedicated to physical training and baseline screening.(Harper et al., 1997) Phase 2 was used to familiarize subjects with the road march course and the performance tasks they were required complete before and after each march.(Harper et al., 1997) Finally, phase 3 consisted of three experimental roach marches, all 10km long with loads of 18, 27 and 36kg.(Harper et al., 1997) All soldiers completed the marches and met the required foot march performance standards provided by U.S. Army doctrine.(Harper et al., 1997) The results showed that men were significantly faster than women in completing maximal effort marches of 10 km under all loading conditions(Harper et al., 1997) Overall, male soldiers completed the marches about 21% faster than the female soldiers.(Harper et al., 1997) Male soldiers maintained a relatively constant pace during the march even though heart rate was seen to increase throughout the march.(Harper et al., 1997) Conversely, female solders maintained a relatively constant heart rate while marching velocity progressively declined throughout the march.(Harper et al., 1997) The results of the study also indicated that males and females reported significantly different ratings of perceived exertion and pain, soreness and discomfort levels.(Harper et al., 1997) Female soldiers reported upper body exertion, and upper and lower back discomfort, significantly higher than male soldiers.(Harper et al., 1997) In addition to the number of sex differences observed, the data show that the load carried by the soldier affects the time to complete a maximal effort road march.(Harper et al., 1997) With heavier loads, subjects decreased the rate of march to maintain heart rates similar to those with lighter loads.(Harper et al., 1997) While heart rates were seen to remain constant between the loading condition, ratings of perceived exertion were seen to increase with increases in load.(Harper et al., 1997) This phenomenon was attributed to ratings of perceived exertion potentially reflecting local muscular strain in addition to cardiopulmonary strain.(Harper et al., 1997) Although female soldiers were able to meet the foot march standards, it should again be noted that these differences may become more pronounced with the recent inclusion of female soldiers into combat arms roles.

2.5 Methodological Considerations

The effects of various perturbations on biomechanical and physiological aspects of gait have been studied in some depth and well documented over the years. Traditionally, gait characteristics are assessed via kinematics, kinetics, and physiological demands. Which instruments to use when assessing gait will depend on the question at hand. When studying gait and trying to capture exactly what is occurring from a global perspective, all assessment tools should be thoroughly reviewed and considered for testing. It is important to bear in mind that gait is a highly complex network of multiply nested subsystems that interrelate to carry out movement. Assessing gait from merely one aspect will not completely capture what is occurring. The current study intends to investigate the effects of load carriage and marching velocity on kinematic and kinetic gait characteristics and will therefore employ laboratory techniques for capturing kinematic and kinetic data.

2.5.1 Kinematic analysis

Gait kinematics are commonly assessed using a 3-deminsional (3D) motion capture system. The *Vicon* motion capture system is a prominent instrument in biomechanics research. A series of reflective markers are placed on various bony landmarks and segments of the body are tracked by a configuration of several cameras to create a 3D model of the body. Data collected from *Vicon* are used to calculate resultant joint angles, joint position, and joint angular velocity and acceleration.

2.5.2 Kinetic analysis

Kinetics of walking and running can be measured by way of force platforms. This is especially useful for load carriage research when looking to assess resultant forces and moments. Force plates are characterized by the number and type of transducers they are equipped with. Some force plates are equipped with only a single transducer (single-pedestal) and others contain multiple transducers (multi-pedestal). Multi-pedestal force plates can measure the forces produced in 3 dimensions (vertical, anteroposterior, mediolateral) and, for that reason, are better suited for gait analysis. Force plates are a staple in biomechanics research, providing the capability to compute forces, moments, directions, torques and centers of pressure.

3.0 Methodology

3.1 Experimental Design

This study will employ a within-subject study design to examine the effect that load carriage mass has on kinematic, kinetic, and spatiotemporal gait characteristics at velocities around the GTP in female subjects. In addition, this study will investigate the impact of load and locomotion on lower limb coordination.

3.1.1 Independent Variables

- Load condition
  - Body weight (BW)
  - Weighted vest loaded with 25% of body weight (+25%BW)
  - Weighted vest loaded with 45% of body weight (+45%BW)
- Marching Velocity (Locomotion)
  - Walk (WK)
  - Run (RN)
  - Forced-march (FM)

3.1.2 Dependent Variables

- Sagittal plane Joint angles for the hip, knee, and ankle (°)

The joint angle formed between the longitudinal axes of adjacent body segments.

- Sagittal plane joint moment at the hip, knee, and ankle (Nm/kg)

  Net internal relative joint moment around the x-lateral axis.

- Stride length (m)

  Distance between proximal end position of the foot at heel-strike to the proximal end position of the foot at the next contralateral heel-strike.

- Stride width (m)

  Medio-lateral distance between proximal end position of the heel-strike to the proximal end position of the foot at the next contralateral heel-strike.

- Stride frequency (strides/min)

  Number of strides per minute.

- Stance time (s)

  Duration of foot contact time from heel-strike to contralateral toe-off.

- Double support time (s)

  Right terminal double limb support plus right initial double limb support.

- Flight time (s)

  Average time from left toe-off to right heel-strike plus the average time form right toe off to left heel-strike.

3.2 Participants

3.2.1 Subject Recruitment

Institutional Review Board approved materials will be used to recruit from the University of Pittsburgh and surrounding institutions intercollegiate athletic and recreationally active populations. Prospective subjects will call the Neuromuscular Research Laboratory at which point they will be asked a series of prescreening questions and subsequently given the opportunity to voluntarily enroll, provided they fulfilled the eligibility requirements.

3.2.2 Subject Consent

Each subject will be thoroughly briefed on all testing procedures and any risks associated with participation. Upon being briefed and provided the opportunity to ask questions, subjects will be asked to sign an informed consent in accordance with the University of Pittsburgh Institutional Review Board before participating in the study.

3.2.3 Inclusions Criteria

To be qualified for the study, participants will meet the following inclusion criteria:

- Female
- Between the ages of 18-30
- Physically active at least 3 days per week, 30-60 minutes per day
- Load carriage experience

3.2.4 Exclusion Criteria

Subjects will be excluded from the study if any of the following exclusion criteria apply:

- Previous diagnosis of conditions requiring additional medical clearance (i.e. cancer, heart disease, or Type I or Type II diabetes)
- Presence of a medical condition that may limit one's ability to bear weight (i.e. orthopedic limitations or severe arthritis).
- Experienced a musculoskeletal injury within the past 6 months that required medical attention
- Known history of any upper or lower extremity injury or surgery
- Frequently experience upper or lower extremity pain with normal activities
- Known history of lower back injury or surgery
- Frequently experience lower back pain with normal activities
- Known history of any other musculoskeletal condition that may affect muscle function or the ability to execute tasks required for the study.
- Currently pregnant

3.3 Sample Size Calculation

An *a priori* power analysis was conducted using G*Power 3.1.9.2 (Heinrich Heine, Universitat Dusseldorf) sample size calculator.(G*Power, 2009) Table 1 depicts the statistical test parameters that were used for the sample size calculation. Given an effect size of 0.20, and an alpha of 0.05, it was estimated that a sample size of 16 would achieve an 81.5% power. To account for a 15% attrition rate and potential data loss, a total of N=18 subjects are needed to participate in this study.

Table 1. Sample size calculation

Statistical Test	Effect Size (f)	Alpha	Power	# of Measures	Corr. Among Measures	Non-sphericity Corr.	Sample Size
ANOVA: Repeated measures, within factor	0.2	0.05	0.8	16	0.5	1	16

3.4 Instrumentation

3.4.1 Anthropometric Measurements

Height (cm) will be measured using a stadiometer (Seca North America; East Hanover, MD). Body weight (kg) will be measured using a scale (BOD POD Version 5.2.0, COSMED USA Inc.; Chicago, IL).

3.4.2 Body Composition

The DEXA (Complete Medical Services, GE Healthcare, Sterling Heights, MI) will be used in the laboratory to measure body composition and bone mineral density. The DEXA utilizes dual x-ray absorptiometry in order to measure body composition (fat, lean, and bone mass) both total body (TB) and regionally. Repeatability of total body DEXA measurements is excellent for bone mineral content (r=0.99), lean mass (r=0.99), fat tissue mass (r=1.00), and bone mineral density (r=0.98) during supine scanning.(Lohman, Harris, Teixeira, & Weiss, 2000)

3.4.3 Three-Dimensional Motion Analysis

A Vicon motion capture system will be used to capture 3-dimensional kinematic data for the assessment of gait characteristics under the various loading conditions. The Vicon motion analysis system uses high-speed infrared cameras to collect three-dimensional trajectory data.(E. A. Harman et al., 2000) These cameras track infrared light that is reflected off retro-reflective markers placed on the subjects' body.(E. A. Harman et al., 2000) For the purpose of this study, retro-reflective markers were placed on the subject's lower extremities and torso. In an effort to mitigate any marker occlusion that resulted from gear placement, twelve cameras were optimally positioned to provide the best visibility to capture marker trajectory data.

3.4.4 Bertec Fully Instrumented Treadmill

The two force plates within the Bertec fully instrumented treadmill will be used to collect ground reaction force data and calculate joint kinetic data. The Bertec treadmill is equipped with two separate running surfaces approximately 60 inches long and 20 inches wide per belt.(Obusek et al., 1997) Each belt is capable of independently measuring force and torque in all six axes.(Obusek et al., 1997) Lower extremity kinetic and kinematic data will be collected simultaneously and time synchronized using Vicon Nexus software. GRF data will be collected in all three planes (vertical, anteroposterior, mediolateral). Analog data from the Bertec treadmill was converted to a digital signal using an analog-to-digital converter.

3.5 Testing Procedures

Subjects will report to the Neuromuscular Research Laboratory (NMRL) on one occasion and the Biodynamics Laboratory (BDL) on two separate occasions for testing. For their first visit, subjects will report to the NMRL for informed consent, anthropometrics, a DEXA scan, and treadmill/load carriage familiarization. Subjects will be asked to wear comfortable fitting athletic attire. Upon arrival to the NMRL, subjects will again be screened for inclusion/exclusion criteria and briefed on the study's purpose and procedures. Subjects will be provided the opportunity to ask questions and voice any concerns. Once the investigator confirmed eligibility and answered any questions, subjects will be asked to sign an informed consent in accordance with the Institutional Review Board requirements. Following informed consent, a urine pregnancy test will be administered. The subjects will be asked to provide a urine sample in a cup which will then be analyzed by one of the co-investigators using a urine pregnancy test stick. After the test results are available and recorded (assuming the results were negative), the subjects height and weight will be measured and a DEXA scan will be performed. Lastly, subjects will complete a load carriage familiarization protocol. The subject will be properly fitted with military style boots and asked to position themselves on the treadmill. The protocol will begin with 45 seconds of walking (3mph) followed by 6 acceleration ramps where the treadmill will accelerate to a slow jogging speed (5mph). Each ramp will last 15 seconds and followed by 30 seconds of walking. This protocol will be performed for each load condition, (BW, +25%BW, +45%BW).

The two subsequent visits will take place at the BDL. Subjects will be asked to arrive at the BDL 15 minutes prior to their scheduled testing time and refrain from engaging in fatiguing physical activity 24 hours prior to their scheduled testing session. Subjects will wear military style boots and a tight fitting lycra suits (spandex, one or two piece) to allow for a more accurate kinematic analysis. Motion capture makers will be bilaterally placed on the subjects 1^{st} and 5^{th} metatarsophalangeal joint (MPJ), calcaneus, medial and lateral malleolus, medial and lateral epicondyle of the femur, greater trochanter of the femur, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), acromion process (anterior aspect), and acromial angle (posterior aspect). An additional marker will be placed on the spine of the 7^{th} cervical vertebra. Lastly, marker clusters will be bilaterally secured to the shank and thigh. Markers on the medial and lateral malleolus, medial and lateral femoral epicondyles, and greater trochanters of the femur are calibration only and will be removed for motion trials.

To establish the GTP, each subject will complete three ramped treadmill protocols for each load condition on the Bertec fully instrumented treadmill immediately prior to data collection for that condition. Mean GTP from the ramped treadmill trials will be calculated for each loading condition and used for the 3 subsequent experimental trials. Once the GTP is established, three trials of data collection will be conducted for each condition using a stepped treadmill protocol. Table 2 and Figure 1 outlines the proceedings for each condition and trial. Each trial will begin with an acceleration ($a=0.05m/s^2$) to a designated velocity. Data collection for the WK trial will begin at a velocity 30% below the subjects GTP and remain at that velocity for one minute before accelerating to a velocity 10% below the GTP for an additional minute. Data collection for the RN trial will follow a similar pattern but begin at a velocity 10% below the GTP and finish at a velocity 10% above the GTP. The FM trial will have the same starting and ending velocities as the RN trial, however, subjects will be instructed to continue walking through the GTP. Each trial will last approximately 2 minutes and 30 seconds, followed by 2-5 minutes of rest to minimize the effects of fatigue. For establishing the GTP and for the WK and RN trials, participants will be given minimal instruction in an effort to elicit a natural/true GTP. Loading conditions will be randomized to minimize any order effects. The entire session will last approximately 1:30 - 2:00 (hh:mm) in duration.

Table 2. Experimental conditions

	BW	+25%BW	+45%BW
Establish GTP (3:00 min)	Ramped Treadmill x 3	Ramped Treadmill x 3	Ramped Treadmill x 3
Trial 1 (2:30 min)	WK	WK	WK
Trial 2 (2:30 min)	RN	RN	RN
Trial 3 (2:30 min)	FM	FM	FM

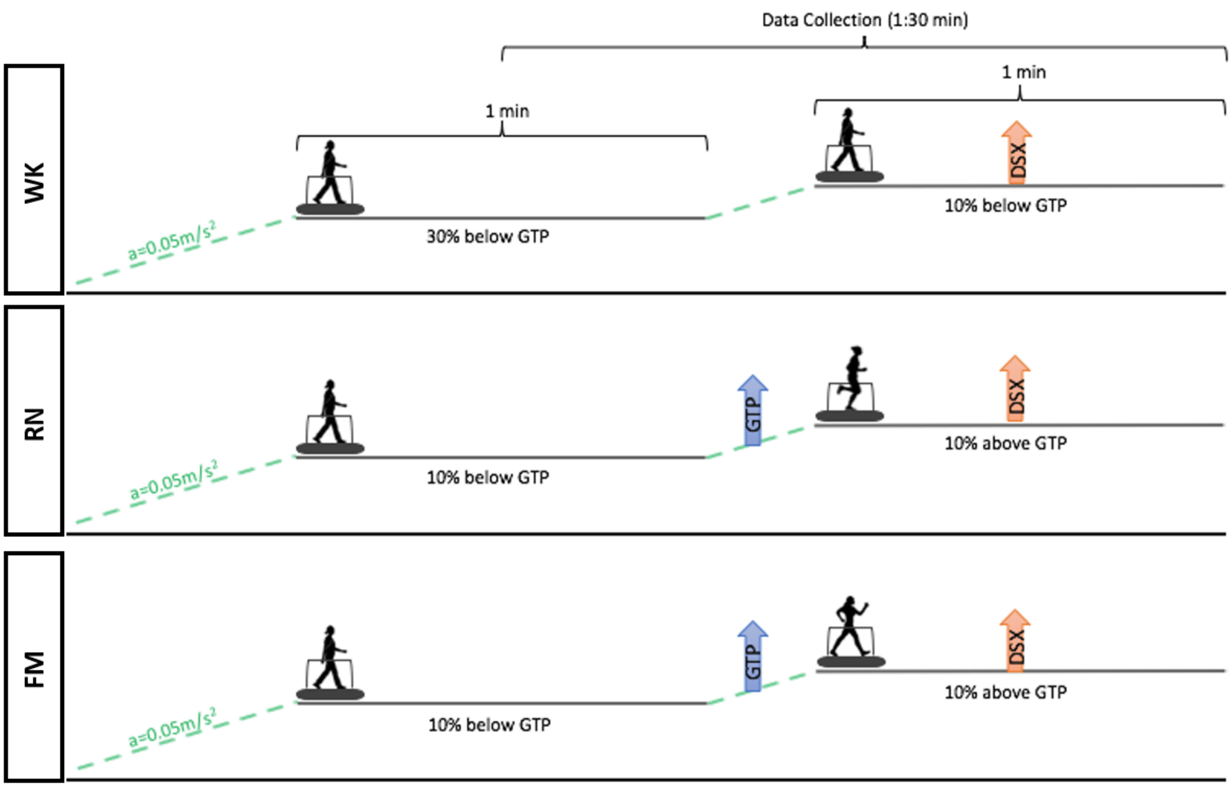

Figure 1. Experimental trials

3.6 Data Reduction

All marker trajectory and ground reaction force data were recorded using the Vicon Nexus 2 software. Each 2.5-minute load carriage trial yielded 90 seconds of force platform and motion capture data. Motion capture data was collected at 100Hz and force plate data at 1000Hz. Using the Vicon Nexus 2 software, a custom labeling template was created for the marker configuration used in the study. Once all static and motion trials were reconstructed, the labeling template was used to auto label the static trials captured for each load condition (BW, +25%BW, +45%BW) which were then used to auto label their respective motion trials (RN, WK, FM). Gap filling methods in Nexus 2 were used to correct any breaks in trajectory data due to marker occlusion. Gaps larger than 20 frames were reviewed first and manually filled using the appropriate fill method deemed to provide the line of best fit. The rigid body fill method was used for any gaps in cluster markers of the thigh and shank segments since those markers were in a fixed position relative to each other. Since all motion trials were conducted on a treadmill and therefore cyclic in nature, an auto gap fill operation using a cyclic fill method was used for any remaining gaps of 20 frames or fewer. Once all motion trials were reconstructed, labeled, and properly gap filled, the trials from testing sessions with the most complete and highest quality data were selected for each subject. From each selected trial, seven smaller segments of data were cropped and exported as C3D files for further processing in Visual3D. These segments included ten complete and consecutive gait cycles at the following time points; pre treadmill acceleration, immediately pre treadmill acceleration, during treadmill acceleration, immediately post treadmill acceleration, post treadmill acceleration, immediately post data collection trigger, and post data collection trigger.

Once all files were imported into Visual3D, trials were grouped by load condition and uploaded together in separate workspaces since all kinetic metrics had to be normalized by total mass. Force platform parameters had to be adjusted for all files to ensure that each force vector was properly assigned to its respective segment. In accordance with the recommended parameters set forth by C-motion when using a Bertec fully instrumented treadmill, the force platform threshold was set to 50N. Any signal value below the threshold was set to zero. All trajectory and analog force platform data was filtered using a fast fourier transform filter with a cutoff frequency of 6 Hz. Net internal moment around the x-lateral, y-anterior, and z-axial axes were calculated for each joint of the lower extremity using inverse dynamics calculations. The segment proximal to the joint was set as the resolution coordinate system and the values were normalized by subject mass (BW, +25%BW, +45%BW). Relative joint angles were derived from Euler angles and absolute segment angles in the right horizontal plane were derived from coplanar angles. Lastly, Continuous relative phase (CRP) was calculated by first constructing phase-plane portraits of normalized angular displacement and angular velocity of each segment. Phase angles were then calculated from the phase-plane trajectories of each segment. Finally, CRP was calculated by subtracting the phase angle of the distal segment (shank) from the phase angle of the proximal segment (thigh) at each instant of time.

3.7 Statistical Analysis

All data was analyzed using SPSS. Descriptive statistics (means, standard deviations, medians, and interquartile ranges as appropriate) were calculated for all variables. Shapiro-Wilk test was used to assess normality for all kinematic, kinetic, and spatiotemporal data. Repeated measures analysis of variance was used to examine the effect of load (3 repeats; BW, +25%BW, +45%BW) and locomotion (3 repeats; WK, RN, FM) on kinematic, kinetic, and spatiotemporal gait characteristics. For data that was not normally distributed, corresponding non-parametric tests were used. Alpha was set *a priori* at 0.05, two-sided. Additionally, variability in gait coordination across load and locomotion conditions was observationally assessed using a CRP analysis.

4.0 Results

The primary purpose of this study was to investigate the main effects of load carriage magnitude on kinematic, kinetic, and spatiotemporal gait characteristics at marching velocities around the GTP in female subjects. As a secondary aim, coordination patterns were examined for each load and locomotion condition to determine levels of variability/stability within the movement system itself.

4.1 Subjects

Due to equipment failure, only 15 female subjects were recruited for the study. One subject withdrew after the familiarization session but before the first experimental testing session. Another subject completed the familiarization session but was unable to complete an experimental testing session as a result of equipment failure. 13 subjects had complete data for processing, however, data from one of the 13 was of very low quality due to severe marker occlusion and the subject's inability to keep each foot on its respective left and right treadmill belt. This subject's data was therefore removed, leaving the data from 12 subjects for the analysis.

Table 3. Subject demographics

Demographic		N	Mean	SD	Min	Max
Age (years)		12	24.75	2.26	21	28
Height (in)		12	64.26	2.55	60.3	67.9
Body Weight (lbs)		12	134.16	21.42	104.8	178.2
Vest Weight (lbs)	+25%BW	12	32.93	5.78	24.2	44.6
	+45%BW	12	58.69	9.79	45.2	77.4
Fat Free Mass (lbs)		12	94.38	14.02	79.8	134.1
Percent Fat Free Mass (%)		12	71.02	8.78	57.5	87.4
Fat Mass (lbs)		12	34.30	15.50	7.8	62.3
Percent Fat Mass (%)		12	25.87	9.56	7.8	40.4

4.2 Spatiotemporal Metrics

3 x 3 two-way repeated measures ANOVA were performed to examine the effects of load (BW, +25%BW, +45%BW) and locomotion (WK, RN, FM) on stride frequency (SF), stride length (SF), stride width (SW), stance time (ST), double support time (DST), flight time (FT), and gait transition point (GTP).

4.2.1 Stride Frequency

SF units are in steps per minute (s/m). There was no significant interaction between load and locomotion, in their effect on SF ($F(4, 44)=1.95$, $p=.119$, $\eta^2_p=.151$). There was no significant main effect of load on SF, averaged across levels of locomotion ($F(2, 22)=1.81$, $p=.188$,$\eta^2_p=.141$). However, there was a significant main effect of locomotion on SF, averaged across levels of load ($F(1.36, 14,96)=121.25$, $p<.001$, $\eta^2_p=.917$). In order to find the pattern of differences on SF depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that SF was significantly different for all locomotion conditions with RN (mean=82.07s/m, se=1.33s/m) eliciting the greatest SF and WK (mean=64.24s/m, se=1.25s/m) the lowest, averaged across levels of load ($p<.001$).

Table 4. Descriptive statistics for stride frequency in steps per minute across all locomotion and load conditions

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	64.03	3.77	63.64	60.18 - 67.29
	+25%BW	12	64.54	4.62	64.96	61.33 - 66.98
	+45%BW	12	64.13	5.11	65.26	59.66 - 67.96
RN [g]	BW	12	81.17	4.34	81.63	78.27 - 83.52
	+25%BW	12	82.88	4.83	82.56	79.89 - 84.80
	+45%BW	12	82.15	5.29	81.60	79.09 - 84.72
FM [g, gg]	BW	12	72.50	5.33	71.38	67.66 - 75.97
	+25%BW	12	74.43	6.26	73.95	69.61 - 77.97
	+45%BW	12	73.97	7.22	75.07	65.89 - 80.16

[w] significantly different from BW [ww] significantly different from +25%BW
[g] significantly different from WK [gg] significantly different from RN

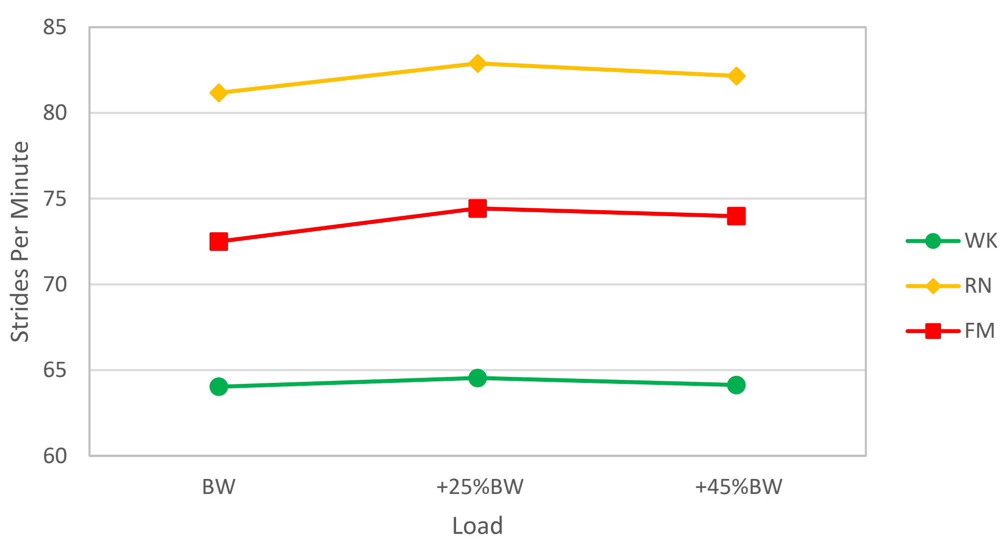

Figure 2. Mean stride frequency across all locomotion and load conditions

4.2.2 Stride Length

SL units are in meters (m). There was no significant interaction between load and locomotion, in their effect on SL ($F(4, 44)=2.52$, $p=.054$, $\eta^2_p=.187$). There was a significant main effect of load on SL, averaged across levels of locomotion ($F(1.27, 13.94)=20.11$, $p<.001$, $\eta^2_p=.646$). In order to find the pattern of differences on SL depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that SL was significantly different for all load conditions with BW (mean=1.58m, se=.038m) eliciting the greatest and +45%BW (mean=1.48m, se=.028m) the lowest SL, averaged across levels of locomotion ($p<.05$).

There was also a significant main effect of locomotion on SL, averaged across load ($F(2, 22)=29.71$, $p<.001$, $\eta^2_p=.730$). In order to find the pattern of differences on SL depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that SL during FM (mean=1.63m, se=.031m) was significantly greater than both WK (mean=1.52m, se=0.30m) and RN (mean=1.46m, se=.048m) locomotion trials, averaged across levels of load ($p<.001$).

Table 5. Descriptive statistics for stride length in meters across all locomotion and load conditions

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	1.56	0.11	1.57	1.54 - 1.62
	+25%BW	12	1.53[w]	0.12	1.53	1.47 - 1.62
	+45%BW	12	1.47[w, ww]	0.09	1.51	1.42 - 1.54
RN [g]	BW	12	1.51	0.18	1.55	1.40 - 1.61
	+25%BW	12	1.47[w]	0.19	1.49	1.34 - 1.61
	+45%BW	12	1.41[w, ww]	0.14	1.44	1.32 - 1.51
FM [g, gg]	BW	12	1.69	0.11	1.71	1.66 - 1.77
	+25%BW	12	1.62[w]	0.12	1.64	1.57 - 1.71
	+45%BW	12	1.56[w, ww]	0.09	1.59	1.49 - 1.63

[w] significantly different from BW [ww] significantly different from +25%BW
[g] significantly different from WK [gg] significantly different from RN

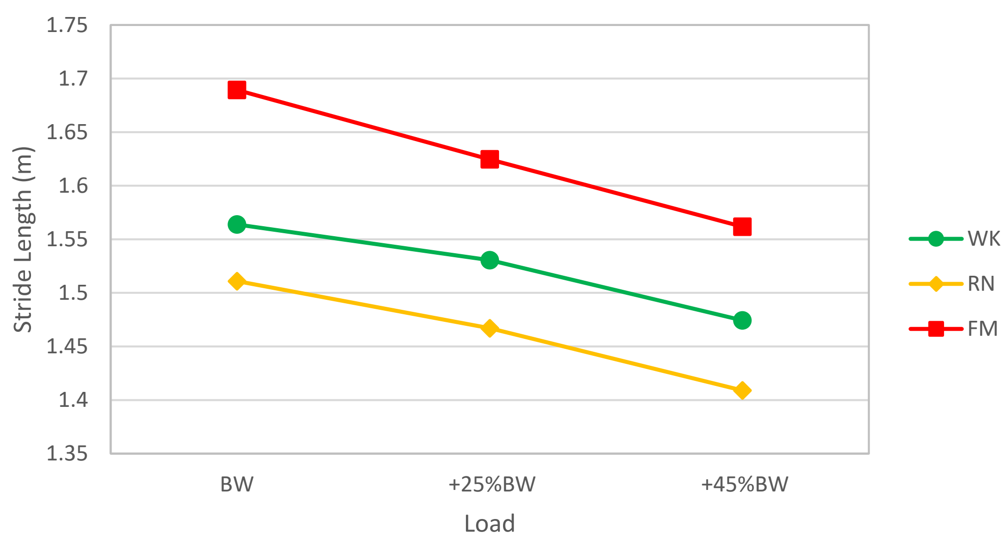

Figure 3. Mean stride length in meters across all locomotion and load conditions

4.2.3 Stride Width

SW units are in meters (m). There was no significant interaction between load and locomotion, in their effect on SF ($F(4, 44)=.21$, $p=.932$, $\eta^2_p=.019$). There was a significant main effect load on SW, averaged across locomotion ($F(2, 22)=7.28$, $p=.004$, $\eta^2_p=.398$). In order to find the pattern of differences on SL depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that SW was significantly greater with +45%BW (mean=.159m, se=.006m) than BW (mean=.149m, se=.006m), averaged across levels of locomotion (p=.011). There was also a significant main effect of locomotion on SW, averaged across load ($F(2, 22)=31.06$, $p<.001$, $\eta^2_p=.738$). In order to find the pattern of differences on SL depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that SW during both WK (mean=.163m, se=.006m) and FM (mean=.163m, se=.006m) was significantly greater than RN (mean=.136m, se=.008m), averaged across levels of load (p<.001).

Table 6. Descriptive statistics for stride width in meters across all locomotion and load conditoins

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	0.15	0.02	0.15	0.14 - 0.17
	+25%BW	12	0.16	0.02	0.16	0.15 - 0.18
	+45%BW	12	0.17^{w}	0.02	0.17	0.15 - 0.18
RN ^{g}	BW	12	0.13	0.03	0.12	0.11 - 0.16
	+25%BW	12	0.13	0.03	0.13	0.12 - 0.16
	+45%BW	12	0.14^{w}	0.02	0.14	0.12 - 0.15
FM ^{gg}	BW	12	0.15	0.02	0.16	0.14 - 0.18
	+25%BW	12	0.16	0.02	0.16	0.15 - 0.17
	+45%BW	12	0.17^{w}	0.03	0.16	0.15 - 0.18

^{w}significantly different from BW ^{ww}significantly different from +25%BW
^{g}significantly different from WK ^{gg}significantly different from RN

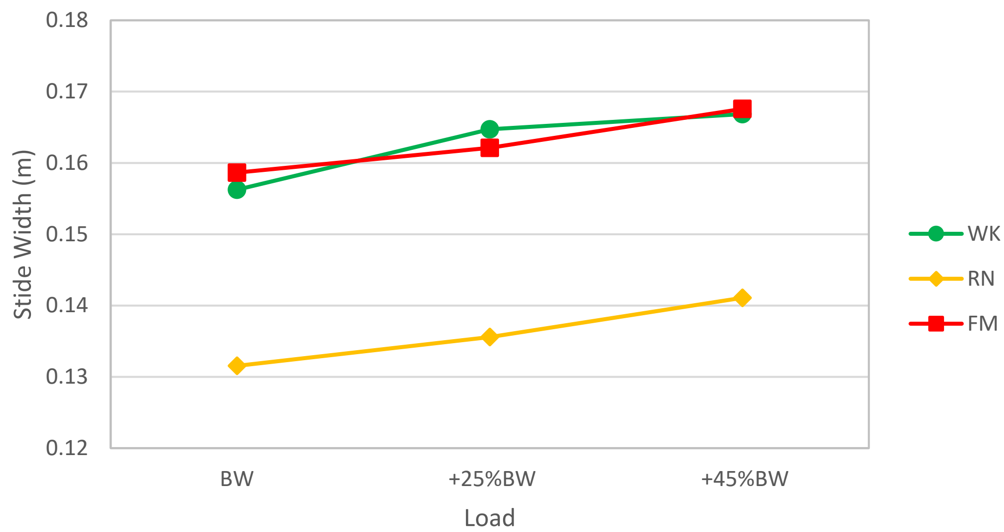

Figure 4. Mean stride width in meters across all locomotion and load conditions

4.2.4 Stance Time

ST units are in seconds (s). There was a significant interaction between the effects of load and locomotion on ST ($F(4, 44)=17.02$, $p<.001$, $\eta^2_p=.61$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22)=8.10$, $p=.002$, $\eta^2_p=.424$). In order to find the pattern of differences on ST depending on load during WK, post hoc marginal pairwise comparisons were performed. Results revealed that ST was significantly greater with +25%BW (mean=.578s, se=.013s) and +45%BW (mean=.592s, se=.015s) when compared to BW (mean=.566s, se=.011s) during WK ($p<.05$). There was also a significant main effect of load during RN ($F(2, 22)=51.38$, $p<.001$, $\eta^2_p=.824$). Pairwise comparisons showed that ST was significantly different for all load conditions with +45%BW (mean=.389s, se=.017s) eliciting the greatest and BW (mean=.314s, se=.013s) the lowest ST during RN ($p<.001$).

Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22)=323.22$, $p<.001, \eta^2_p=.967$), +25%BW ($F(2, 22)=298.29$, $p<.001, \eta^2_p=.964$), +45%BW ($F(1.26, 13.86)=291.40$, $p<.001, \eta^2_p=.964$)]. To identify the pattern of differences on ST depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on ST were similar at each level of load. Results revealed that ST was significantly different for all locomotion patterns with WK [BW (mean=.566s, se=.011s), +25%BW (mean=.578s, se=.013s), +45%BW (mean=.592s, se=.015s)] eliciting the greatest and RN [BW (mean=.314s, se=.013s), +25%BW (mean=.344s, se=.013s), +45%BW (mean=.389s, se=.017s)] the lowest ST at each level of load ($p<.001$).

Table 7. Descriptive statistics for stance time in seconds across all locomotion and load conditions

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	0.56	0.04	0.57	0.52 - 0.60
	+25%BW	12	0.57^{w}	0.04	0.57	0.55 - 0.60
	+45%BW	12	0.59^{w}	0.05	0.59	0.54 - 0.63
RN ^{g}	BW	12	0.31	0.04	0.31	0.30 - 0.33
	+25%BW	12	0.34^{w}	0.05	0.34	0.32 - 0.37
	+45%BW	12	0.39^{w}	0.06	0.39	0.36 - 0.44
FM $^{g, gg}$	BW	12	0.50	0.04	0.50	0.48 - 0.53
	+25%BW	12	0.50^{w}	0.04	0.51	0.47 - 0.54
	+45%BW	12	0.52^{w}	0.05	0.51	0.47 - 0.57

^{w} significantly different from BW ^{ww} significantly different from +25%BW
^{g} significantly different from WK ^{gg} significantly different from RN

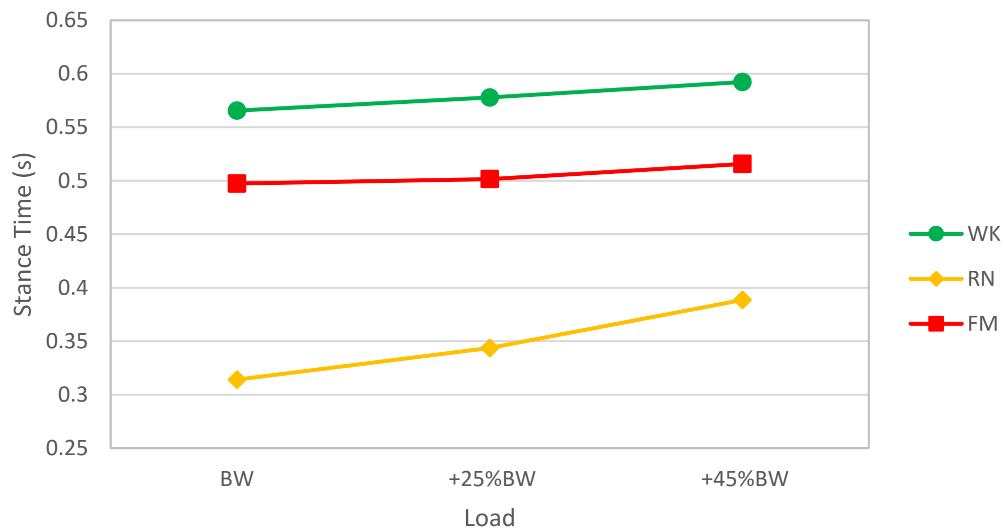

Figure 5. Mean stance time in seconds across all locomotion and load conditions

4.2.5 Double Support Time

DST units are in seconds (s). There was no significant interaction between load and locomotion, in their effect on DST ($F(2, 22)=1.33$, $p=.286$, $\eta^2_p=.108$). There was a significant main effect of load on DST, averaged across WK and FM locomotion trials ($F(2, 22)=52.37$, $p<.001$, $\eta^2_p=.826$). In order to find the pattern of differences on DST depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that DST was significantly different for all load conditions, with +45%BW (mean=.230s, se=.010s) eliciting the greatest and BW (mean=.182s, se=.005s) the lowest DST, averaged across WK and FM locomotion trials ($p\leq.004$).

Additionally, there was a significant main effect of locomotion on DST, averaged across load ($F(1, 11)=22.24$, $p<.001$, $\eta^2_p=.669$). In order to find the pattern of differences on DST depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that DST during WK (mean=.220s, se=.008s) was significantly higher than during FM (mean=.194s, se=.006s), averaged across levels of load (p=.001).

Table 8. Descriptive statistics for double support time in seconds during walk and forced march conditions across all load conditions

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	0.19	0.03	0.20	0.17 - 0.20
	+25%BW	12	0.22[w]	0.03	0.22	0.20 - 0.24
	+45%BW	12	0.25[w, ww]	0.04	0.25	0.22 - 0.27
FM [g]	BW	12	0.17	0.01	0.17	0.17 - 0.18
	+25%BW	12	0.20[w]	0.03	0.20	0.18 - 0.22
	+45%BW	12	0.21[w, ww]	0.03	0.21	0.18 - 0.24

[w] significantly different from BW [ww] significantly different from +25%BW
[g] significantly different from WK [gg] significantly different from RN

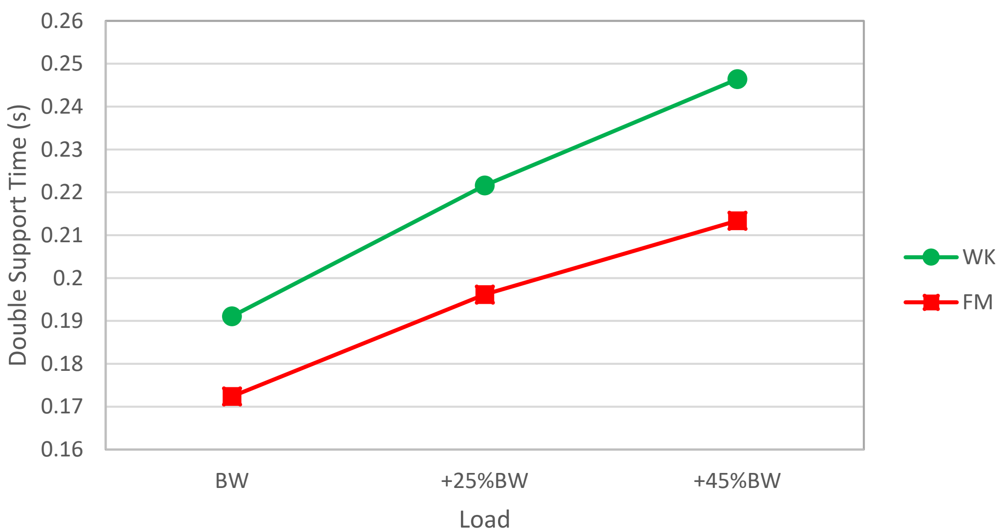

Figure 6. Mean double support time in seconds during walk and forced march conditions across all load conditions

4.2.6 Flight Time

FT units are in seconds (s). There was a significant main effect of load on FT during RN ($F(1.33, 14.63)=50.55$, $p<.001$, $\eta^2_p=.821$). In order to identify the pattern of differences on FT depending on load during RN, post hoc marginal pairwise comparisons were performed. Results showed that FT was significantly different for all load conditions, with BW (mean=.059s, se=.009s) eliciting the greatest and +45%BW (mean=.009s, se=.005s) the lowest FT during RN (p.<001).

Table 9. Descriptive statistics for flight time in seconds while running for each load condition

Locomotion	Load	N	Mean	SD	Med	IQR
RN	BW	12	0.06	0.03	0.07	0.04 - 0.07
	+25%BW	12	0.03^{w}	0.02	0.02	0.02 - 0.03
	+45%BW	12	0.01$^{w, ww}$	0.02	0.00	0.00 - 0.02

^{w}significantly different from BW ^{ww}significantly different from +25%BW

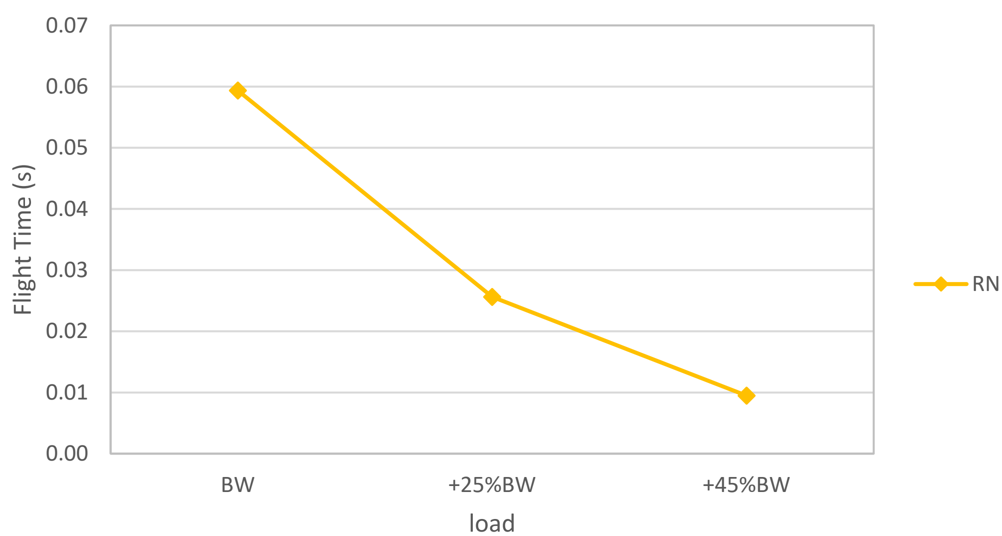

Figure 7. Mean flight time in seconds while running for each load condition

4.2.7 Gait Transition Point

GTP velocity units are in meters per second (m/s). There was a significant main effect of load on GTP ($F(2, 22)=6.98$, $p=.005$, $\eta^2_p=.388$). In order to identify the pattern of differences on FT depending on load during RN, post hoc marginal pairwise comparisons were performed. Results showed that the GTP velocity at BW (mean=1.86m/s, se=.056m/s) and +25%BW (mean=1.83m/s, se=.065m/s) was significantly higher than with the +45%BW (mean=1.75m/s, se=.055m/s) load condition ($p<.029$). There was no significant difference between BW and +25%BW conditions.

Table 10. Descriptive statistics for gait trainsition point in meters per second for each load condition

Load	N	Mean	SD	Med	IQR
BW	12	1.86	0.19	1.83	1.76 - 1.98
+25%BW	12	1.83	0.22	1.82	1.69 - 2.01
+45%BW	12	1.75[w]	0.19	1.75	1.67 - 1.87

[w] significantly different from BW [ww] significantly different from +25%BW

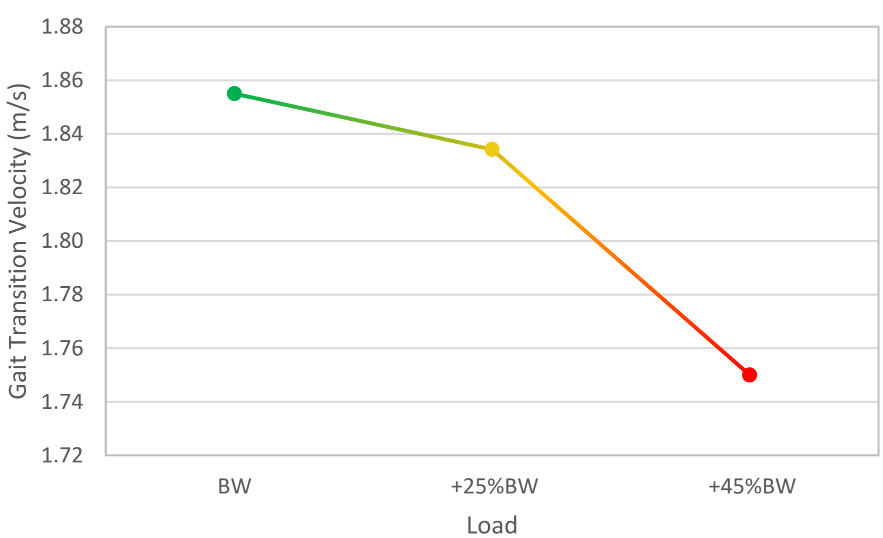

Figure 8. Mean gait transition point velocity in meters per second for each load conditoin

4.3 Left Leg Kinematics

3 x 3 Two-way repeated measures ANOVA were performed to examine the effects of load (BW, +25%BW, +45%BW) and locomotion (WK, RN, FM) on left hip (LHP), knee (LKN) and ankle (LANK) angles at heel strike (HS), mid stance (MS), and toe off (TO). Results reported below are grouped by gait event and ordered by joint from superior to inferior. Absolute joint angles are reported and the units are in degrees (°). Lower values represent a more flexed joint position (dorsiflexion at the ankle) and higher values represent a more extended joint position (plantarflexion at the ankle). **Table 10** and **Figure 7** in this section depict all the results for left leg kinematics.

4.3.1 Heel Strike

There was no significant interaction between load and locomotion, in their effect on HS LHP angle ($F(4, 44) = .69$, $p = .602$, $\eta_p^2 = .059$). There was no significant main effect of load on HS LHP angle, averaged across levels of locomotion ($F(2, 22) = 2.42$, $p = .112$, $\eta_p^2 = .180$). There was, however, a significant main effect of locomotion on HS LHP angle, averaged across levels of load ($F(1.34, 14.70) = 15.40$, $p = .001$, $\eta^2_p = .583$). In order to find the pattern of differences on HS LHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that HS LHP angle was significantly greater during WK (mean =167.05°, se =7.72°) and RN (mean =165.10°, se =2.63°) when compared to FM (mean =161.43°, se =2.87°), averaged across levels of load ($p \leq .004$).

There was no significant interaction between load and locomotion, in their effect on HS LKN angle ($F(4, 44) = .57$, $p = .685$, $\eta_p^2 = .049$). There was a significant main effect of load on HS

LKN angle, averaged across locomotion ($F(2, 22) = 30.62$, $p < .001$, $\eta^2_p = .736$). In order to find the pattern of differences on HS LKN angle depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that HS LKN angle was significantly greater at BW (mean =156.59°, se =1.35°) when compared to +25%BW (mean =153.22°, se =1.53°) and +45%BW (mean =151.30°, se =1.62°) load conditions, averaged across levels of locomotion ($p \leq .001$). Additionally, there was a significant main effect of locomotion on HS LKN angle, averaged across load ($F(2, 22) = 58.97$, $p < .001$, $\eta^2_p = .843$). Again, post hoc marginal pairwise comparisons using the Bonferroni adjustment were performed to reveal the pattern of differences on HS LKN angle depending on locomotion, averaged across load. Results showed that HS LKN angle was significantly different for all locomotion trials, with WK (mean =161.57°, se =1.71°) eliciting the greatest HS LKN angle and RN (mean =146.63°, se =1.42°) the lowest, averaged across levels of load ($p \leq .001$).

There was a significant interaction between the effects of load and locomotion on HS LANK angle ($F(1.88, 20.72) = 4.18$, $p = .032$, $\eta^2_p = .275$). Simple main effects of load were analyzed at each level of locomotion. There was no significant main effect of load at any level of locomotion. Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(1.19, 13.10) = 81.71$, $p < .001$, $\eta^2_p = .881$), +25%BW ($F(2, 22) = 127.08$, $p < .001$, $\eta^2_p = .920$), +45%BW ($F(2, 22) = 91.67$, $p < .001$, $\eta^2_p = .893$)]. To identify the pattern of differences on HS LANK angle depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on HS LANK angle were similar at each level of load. Results revealed that HS LANK angle was significantly different for all locomotion trials with WK [BW (mean=121.25°, se=1.26°), +25%BW (mean=121.90°, se=1.85°), +45%BW (mean=119.89°, se=1.30°)] eliciting the greatest HS LANK angle and RN [BW (mean=109.05°, se=1.62°), +25%BW (mean=110.59°, se=1.80°), +45%BW (mean=109.95°, se=1.26°)] the lowest at each level of load (P<.001).

4.3.2 Mid Stance

There was no significant interaction between load and locomotion, in their effect on MS LHP angle ($F(4, 44) = 1.60$, $p = .192$, $\eta_p^2 = .127$). There was no significant main effect of load on MS LHP angle, averaged across levels of locomotion ($F(2, 22) = .543$, $p = .589$, $\eta_p^2 = .047$). There was, however, a significant main effect of locomotion on MS LHP angle, averaged across levels of load ($F(2, 22)=61.98$, $p<.001$, $\eta^2_p = .849$). In order to find the pattern of differences on MS LHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that MS LHP angle was significantly different for all locomotion trials, with WK (mean =202.38°, se =2.54°) eliciting the greatest MS LHP angle and RN (mean =190.81°, se =2.32°) the lowest, averaged across levels of load ($p \leq .19$).

There was no significant interaction between load and locomotion, in their effect on MS LKN angle ($F(2.72, 29.97) = 1.25$, $p = .305$, $\eta_p^2 = .102$). There was a significant main effect of load MS LKN angle, averaged across levels of locomotion ($F(2, 22) = 10.91$, $p=.001$, $\eta^2_p = .498$). In order to find the pattern of differences on MS LKN angle depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that MS LKN angle was significantly greater at BW (mean =170.71°, se =1.32°) when compared to +25%BW (mean =168.40°, se =1.28°) and +45%BW (mean =167.28°, se =1.35°) load conditions, averaged across levels of locomotion ($p<.05$). Additionally, there was a significant main effect of locomotion on MS LKN angle, averaged across levels of load ($F(1.34, 14.70)=101.76$, $p<.001$, $\eta^2_p =.902$) on MS LKN angle. Again, post hoc marginal pairwise comparisons using the Bonferroni adjustment were performed to reveal the pattern of differences on MS LKN angle depending on locomotion, averaged across load. Results showed that MS LKN angle was significantly greater during WK (mean =175.30°, se =1.05°) and FM (mean =173.34°, se =1.92°) when compared to RN (mean =157.74°, se =1.29°) locomotion trials, averaged across levels of load($p<.001$).

There was a significant interaction between the effect of load and locomotion on MS LANK angle ($F(4, 44)=3.43$, $p=.016$, $\eta^2_p =.238$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) =6,85$, $p =.005$, $\eta_p^2 =.384$) In order to find the pattern of differences on MS LANK angle depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that MS LANK angle was significantly greater at BW (mean =113.28°, se =1.27°) and +25%BW (mean =114.05°, se =1.84°) when compared to +45%BW (mean =111.40°, se =1.32) load condition during WK ($p\leq.13$). Similar results were seen in the effects of load during RN ($F(2, 22) =6.71$, $p =.005$, $\eta_p^2 =.379$) Pairwise comparisons showed that MS LANK angle was significantly greater at BW (mean =108.31°, se =1.48°) and +25%BW (mean =107.29°, se =2.09°) when compared to +45%BW (mean =105.02°, se =1.39°) load condition during RN ($p\leq.39$). There was also a significant main effect of load during FM ($F(2, 22) =4.81$, $p =.019$, $\eta_p^2 =.304$). Pairwise comparisons revealed that MS LANK angle was significantly greater at BW (mean =114.24°, se =1.29°) when compared to +45%BW (mean =111.41°, se =1.64°) load condition during FM ($p=.007$). Additionally, simple main of locomotion on MS LANK angle were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW (F(2, 22)=24.63, p<.001, η^2_p=.691), +25%BW (F(2, 22)=41.22, p<.001, η^2_p=.789), +45%BW (F(2, 22)=34.89, p<.001,η^2_p=760)]. To identify the pattern of differences on MS LANK angle depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on MS LANK angle were similar at each level of load. Results revealed that MS LANK angle was significantly greater during WK [BW (mean=113.28°, se=1.27°), +25%BW (mean=114.05°, se=1.84°), +45%BW (mean=111.40°, se=1.32°)] and FM [BW (mean=114.24°, se=1.29°), +25%BW (mean=113.53°, se=2.04°), +45%BW (mean=111.41°, se=1.64°)]when compared to RN [BW (mean=108.31°, se=1.48°), +25%BW (mean=107.29°, se=2.09°), +45%BW (mean=105.02°, se=1.39)] at each level of load (P<.001).

4.3.3 Toe Off

There was no significant interaction between load and locomotion, in their effect on TO LHP angle (F(4, 44) =.85 p =.505, η_p^2 =.071).There was no significant main effect of load on TO LHP angle, averaged across levels of locomotion (*F*(2, 22) =2.72, *p* =.088, η_p^2 =.198). However, there was a significant main effect of locomotion on TO LHP angle, averaged across levels of load (F(2, 22) =27.74, p<.001, η^2_p =.716). In order to find the pattern of differences on TO LHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that TO LHP angle was significantly greater during WK (mean =185.52°, se =2.26°) when compared to both RN (mean =176.32°, se=2.25°) and FM (mean =175.77°, se =2,15°) locomotion trials, averaged across levels of load (p≤.001).

There was a significant interaction between the effect of load and locomotion on TO LKN angle (F(4, 44)=4.94, p=.002, η^2_p =.310). Simple main effects of load were analyzed for each level of locomotion. There was a significant main effect of load at each level of locomotion [WK ($F(2, 22)=26.58$, $p<.001$, $\eta^2_p=.707$), RN ($F(2, 22)=37.52$, $p<.001$, $\eta^2_p=.773$), FM ($F(2, 22)=26.97$, $p<.001$, $\eta^2_p=.710$)]. To identify the pattern of differences on TO LKN angle depending on load, post hoc marginal pairwise comparisons were performed. The effects of load on TO LKN angle were similar at each level of locomotion. Results revealed that TO LKN angle was significantly different for all load conditions, with BW [WK (mean=148.23°, se=1.25°), RN (mean=146.12°, se=1.65°), FM (mean=148.01°, se=1.21°)] eliciting the greatest TO LKN angle and +45%BW [WK (mean=144.16°, se=1.23°), RN (mean=138.62°, se=1.26°), FM (mean=142.53°, se=1.31°)] the lowest at each level of locomotion ($p\leq.011$). Additionally, simple main effects of locomotion on TO LKN angle were analyzed for each level of load. There was a significant main effect of locomotion with +25%BW ($F(2, 22)=6.90$, $p=.005$, $\eta^2_p=.385$). To identify the pattern of differences on TO LKN angle depending on locomotion with +25%BW, post hoc marginal pairwise comparisons were performed. Results showed that TO LKN angle was significantly greater during WK (mean=146.82°, se=1.24°) when compared to RN (mean=141.88°, se=1.45°) with +25%BW ($p=.002$). There was also a significant main effect of locomotion with +45%BW ($F(2, 22)=11.35$, $p<.001$, $\eta^2_p=.508$). Pairwise comparisons showed that TO LKN angle was significantly greater during both WK (mean=144.16°, se=1.23°) and FM (mean=142.53°, se=1.31°) when compared to RN (mean=138.62°, se=1.26°) with +45%BW ($p\leq.011$).

There was a significant interaction between the effect of load and locomotion on TO LANK angle ($F(4, 44)=8.02$, $p<.001$, $\eta^2_p=.422$). Simple main effects of load were analyzed at each level of locomotion. There was no significant main effect of load at any level of locomotion. Additionally, simple main effects of locomotion on TO LANK angle were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(1.26, 13.82)=87.47$, $p<.001$,

η^2_p=.888). To identify the pattern of differences on TO LANK angle depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results showed that TO LANK angle was significantly different for all locomotion trials, with WK (mean=122.32°, se=1.27°) eliciting the greatest TO LANK angle and RN (mean=113.21°, se=1.48°) the lowest at BW ($p \leq .029$). There was also a significant main effect of locomotion with +25%BW ($F(2, 22)=151.35$, $p<.001$, η^2_p=.932). Pairwise comparisons revealed that TO LANK angle was significantly greater during both WK (mean=122.82°, se=1.90°) and FM (mean=122.09°, se=2.08°) when compared to RN (mean=114.65°, se=1.88°) with +25%BW ($p<.001$). Lastly, there was a significant main effect of locomotion with +45%BW ($F(2, 22)=87.99$, $p<.001$, η^2_p=.889). Pairwise comparisons showed that TO LANK angle was significantly different for all locomotion trials, with WK (mean=121.26°, se=1.38°) eliciting the greatest TO LANK angle and RN (mean114.32°, se=1.39°) the lowest with +45%BW ($p \leq .03$).

Table 11. Descriptive statistics for left lower limb joint angles in degrees at heel strike, mid stance and toe off across all load and locomotion conditions

HIP

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	168.60	9.81	169.36	158.37 - 176.50
	25%BW	12	165.23	10.55	169.86	153.07 - 173.87
	45%BW	12	167.30	10.03	167.04	157.49 - 173.14
RN	BW	12	167.46	9.08	170.47	158.14 - 174.99
	25%BW	12	162.98	10.02	164.49	150.90 - 171.87
	45%BW	12	164.87	10.27	166.37	154.63 - 171.39
FM [g, gg]	BW	12	163.25	10.76	164.82	152.44 - 171.67
	25%BW	12	160.09	11.40	163.10	148.57 - 170.07
	45%BW	12	160.94	9.26	162.02	151.82 - 169.45

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	202.59	10.65	200.93	196.62 - 211.81
	25%BW	12	202.04	9.92	199.91	195.67 - 209.02
	45%BW	12	202.50	7.70	203.21	198.27 - 207.92
RN [g]	BW	12	190.83	7.98	189.93	185.81 - 196.20
	25%BW	12	189.60	8.17	191.18	183.20 - 194.73
	45%BW	12	191.99	8.64	194.69	186.62 - 197.27
FM [g, gg]	BW	12	199.83	10.30	199.33	195.41 - 202.64
	25%BW	12	196.99	9.02	195.95	189.50 - 203.74
	45%BW	12	198.03	7.23	198.52	192.74 - 204.20

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	183.76	9.07	181.67	178.81 - 188.75
	25%BW	12	181.39	8.02	181.19	174.54 - 186.61
	45%BW	12	182.42	8.18	182.41	176.28 - 185.95
RN [g]	BW	12	177.99	7.94	178.82	170.00 - 184.69
	25%BW	12	174.82	8.19	177.53	166.14 - 182.01
	45%BW	12	176.16	8.60	178.19	169.76 - 180.42
FM [g]	BW	12	177.97	8.63	179.24	170.44 - 181.11
	25%BW	12	174.32	8.17	175.12	165.95 - 181.23
	45%BW	12	175.02	7.06	175.22	170.05 - 180.23

KNEE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	164.13	4.80	163.71	159.98 - 166.53
	25%BW	12	161.27 [w]	6.52	160.37	156.62 - 163.18
	45%BW	12	159.33 [w]	7.45	158.62	154.59 - 163.92
RN [g]	BW	12	149.28	5.35	149.32	145.72 - 154.74
	25%BW	12	146.20 [w]	5.14	145.77	143.16 - 148.21
	45%BW	12	144.41 [w]	5.01	144.09	139.49 - 148.18
FM [g, gg]	BW	12	156.38	6.39	157.57	150.45 - 162.80
	25%BW	12	152.19 [w]	6.89	152.12	147.22 - 157.74
	45%BW	12	150.16 [w]	6.38	149.27	146.60 - 155.35

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK [gg]	BW	12	176.61	3.69	177.79	173.75 - 179.50
	25%BW	12	175.60 [w]	3.87	177.06	172.29 - 178.58
	45%BW	12	173.70 [w]	4.26	173.42	171.71 - 176.59
RN [g]	BW	12	159.89	5.44	160.65	155.57 - 163.95
	25%BW	12	157.12 [w]	4.90	158.12	152.73 - 160.81
	45%BW	12	156.23 [w]	4.58	156.87	151.83 - 161.30
FM [gg]	BW	12	175.63	6.43	176.47	168.40 - 181.33
	25%BW	12	172.47 [w]	6.67	173.94	165.38 - 178.16
	45%BW	12	171.91 [w]	7.66	170.34	167.30 - 178.70

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	148.23	4.33	148.55	145.71 - 151.54
	25%BW	12	146.82 [w]	4.29	146.48	144.21 - 150.72
	45%BW [gg]	12	144.16 [w,ww]	4.26	144.86	141.02 - 147.24
RN	BW	12	146.12	5.71	147.57	142.05 - 150.55
	25%BW [g]	12	141.88 [w]	5.03	143.13	139.36 - 145.84
	45%BW [g]	12	138.62 [w,ww]	4.35	137.93	135.67 - 142.43
FM	BW	12	148.01	4.18	148.97	143.99 - 151.16
	25%BW	12	144.55 [w]	4.54	144.24	141.48 - 148.78
	45%BW [gg]	12	142.53 [w,ww]	4.54	142.34	138.44 - 146.37

ANKLE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	121.35	4.37	122.12	117.20 - 124.91
	25%BW	12	121.90	6.44	123.62	114.39 - 126.42
	45%BW	12	119.89	4.50	120.43	115.67 - 123.65
RN [g]	BW	12	109.05	5.62	110.28	103.78 - 112.37
	25%BW	12	110.59	6.23	112.55	106.40 - 115.09
	45%BW	12	109.95	4.38	110.28	108.92 - 113.29
FM [g, gg]	BW	12	118.41	4.91	119.86	112.45 - 122.33
	25%BW	12	118.50	7.46	120.73	111.93 - 123.17
	45%BW	12	117.06	5.38	118.71	112.88 - 121.08

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK [gg]	BW	12	113.28	4.40	114.62	109.52 - 116.57
	25%BW	12	114.05	6.38	116.32	108.04 - 119.28
	45%BW	12	111.40 [w, ww]	4.56	112.15	107.06 - 114.28
RN [g]	BW	12	108.31	5.14	108.91	102.93 - 111.21
	25%BW	12	107.29	7.24	108.64	101.11 - 113.17
	45%BW	12	105.02 [w, ww]	4.83	106.47	102.79 - 107.77
FM [gg]	BW	12	114.24	4.46	114.95	110.87 - 118.17
	25%BW	12	113.53	7.07	114.53	105.46 - 120.47
	45%BW	12	111.40 [w]	5.68	111.94	106.92 - 116.52

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	122.32	4.39	124.12	118.05 - 124.50
	25%BW [gg]	12	122.82	6.58	125.35	116.85 - 128.23
	45%BW	12	121.26	4.76	122.12	118.94 - 124.15
RN	BW [g]	12	113.21	5.13	114.59	108.15 - 117.59
	25%BW [g]	12	114.65	6.50	116.23	109.71 - 119.29
	45%BW [g]	12	114.32	4.82	115.89	112.10 - 118.02
FM	BW [g, gg]	12	121.28	4.78	123.66	115.88 - 124.70
	25%BW [gg]	12	122.09	7.20	124.23	114.58 - 128.09
	45%BW [g, gg]	12	120.16	5.36	121.54	117.11 - 123.47

[w] significantly different from BW [ww] significantly different from +25%BW [g] significantly different from WK [gg] significantly different from RN

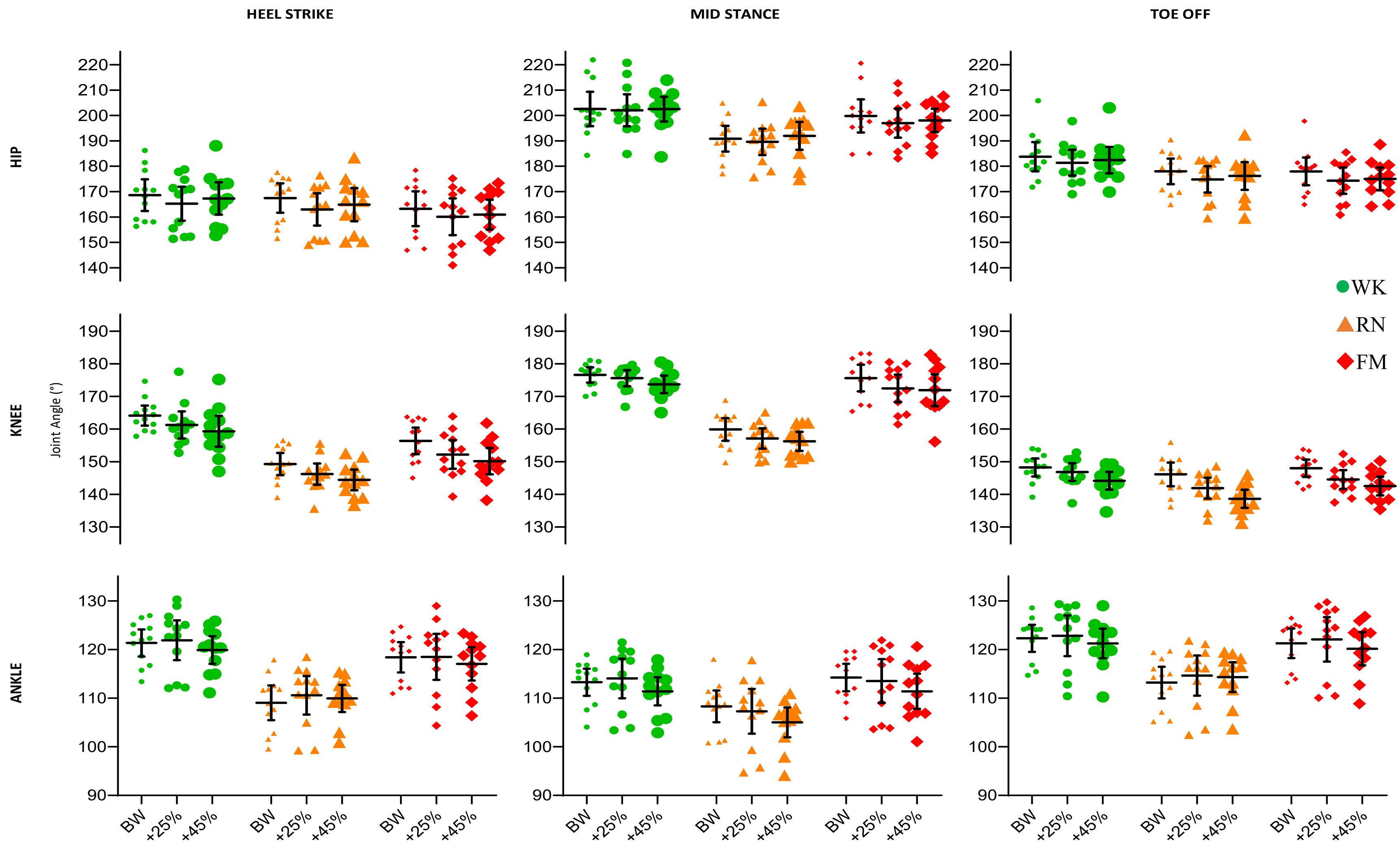

Figure 9. Left lower limb mean joint angles in degrees with 95% CI at heel strike, mid stance and toe off across all load and locomotion conditions

4.4 Right Leg Kinematics

3 x 3 two-way repeated measures ANOVA were conducted to examine the effects of load (BW, +25%BW, +45%BW) and locomotion (WK, RN, FM) on right hip (RHP), knee (RKN) and ankle (RANK) angles at heel strike (HS), mid stance (MS), and toe off (TO). Results reported below are grouped by gait event and ordered by joint from superior to inferior. Absolute joint angles are reported and the units are in degrees (°). Lower values represent a more flexed joint position (dorsiflexion at the ankle) and higher values represent a more extended joint position (plantarflexion at the ankle). **Table 11** and **Figure 8** in this section depict all the results for right leg kinematics.

4.4.1 Heel Strike

There was no significant interaction between load and locomotion, in their effect on HS RHP angle ($F(4, 44) = 1.01$, $p = .413$, $\eta_p^2 = .084$). There was no significant main effect of load on HS RHP angle, averaged across levels of locomotion ($F(2, 22) = .07$, $p = .936$, $\eta_p^2 = .006$). There was, however, a significant main effect of locomotion on HS RHP angle, averaged across levels of load ($F(2,22) = 13.48$, $p < .001$, $\eta^2_p = .551$). In order to find the pattern of differences on HS RHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that HS RHP angle was significantly greater during WK (mean =167.68°, se =2.91°) and RN (mean =166.05°, se =3.23°) when compared to FM (mean =162.00°, se =3.29°), averaged across levels of load ($p \leq .008$).

There was no significant interaction between load and locomotion, in their effect on HS RKN angle ($F(4, 44) =.285$, $p =.886$, $\eta_p^2 =.025$).There was a significant main effect of load on HS RKN angle, averaged across locomotion ($F(2, 22)= 27.64$, $p<.001$, $\eta^2_p =.715$). In order to find the pattern of differences on HS RKN angle depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that HS RKN angle was significantly greater at BW (mean =156.73°, se =2.01°) when compared to +25%BW (mean =153.14°, se =2.23°) and +45%BW (mean =151.73°, se =2.10°) load conditions, averaged across levels of locomotion ($p\leq.002$). Additionally, there was a significant main effect of locomotion on HS RKN angle, averaged across load ($F(2, 22) =48.68$, $p<.001$, $\eta^2_p =.816$). Again, post hoc marginal pairwise comparisons using the Bonferroni adjustment were performed to reveal the pattern of differences on HS RKN angle depending on locomotion, averaged across load. Results showed that HS RKN angle was significantly different for all locomotion trials, with WK (mean =161.33°, se =2.12°) eliciting the greatest HS RKN angle and RN (mean =147.30°, se =2.06°) the lowest, averaged across levels of load ($p\leq.006$).

There was a significant interaction between the effects of load and locomotion on HS RANK angle ($F(4, 44)= 8.02$, $p<.001$, $\eta^2_p =.422$). Simple main effects of load were analyzed at each level of locomotion. There was no significant main effect of load on HS RANK angle at any level of locomotion. Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22)=67.76$, $p<.001$, $\eta^2_p=.860$), +25%BW ($F(2, 22)=113.202$, $p<.001$, $\eta^2_p=.911$), +45%BW ($F(2, 22)=100.70$, $p<.001$, $\eta^2_p=.902$)]. To identify the pattern of differences on HS RANK angle depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on HS RANK angle were similar at each level of load.

Results revealed that HS RANK angle was significantly different for all locomotion trials with WK [BW (mean=120.16°, se=1.25°), +25%BW (mean=120.16°, se=1.18°), +45%BW (mean=118.88°, se=1.42°)] eliciting the greatest HS RANK angle and RN [BW (mean=108.12°, se=1.62°), +25%BW (mean=109.40°, se=1.47°), +45%BW (mean=110.24°, se=1.56°)] the lowest at each level of load ($p \leq .001$).

4.4.2 Mid Stance

There was no significant interaction between load and locomotion, in their effect on MS RHP angle ($F(=4, 44) = 1.98$, $p = .115$, $\eta_p^2 = .152$).There was no significant main effect of load on MS RHP angle, averaged across levels of locomotion ($F(2, 22) = .384$, $p = .686$, $\eta_p^2 = .034$). There was, however, a significant main effect of locomotion on MS RHP angle, averaged across levels of load ($F(2, 22) = 30.48$, $p < .001$, $\eta^2_p = .735$). In order to find the pattern of differences on MS RHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that MS RHP angle was significantly different for all locomotion trials, with WK (mean =202.08°, se =2.05°) eliciting the greatest MS RHP angle and RN (mean =192.46°, se =2.30°) the lowest, averaged across levels of load ($p \leq .028$).

There was no significant interaction between load and locomotion, in their effect on MS RKN angle ($F(=4, 44) = .563$, $p = .691$, $\eta_p^2 = .049$). There was a significant main effect of load MS RKN angle, averaged across levels of locomotion ($F(2, 22) = 6.31$, $p = .007$, $\eta^2_p = .365$). In order to find the pattern of differences on MS RKN angle depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that MS RKN angle was significantly greater at BW (mean = 169.91°, se =1.58°) when compared to +45%BW (mean =167.61°, se =1.48°) load condition, averaged across levels of locomotion ($p=.023$). Additionally, there was a significant main effect of locomotion on MS RKN angle, averaged across levels of load ($F(2, 22)= 55.71$, $p<.001$, $\eta^2_p =.835$) on MS RKN angle. Again, post hoc marginal pairwise comparisons using the Bonferroni adjustment were performed to reveal the pattern of differences on MS RKN angle depending on locomotion, averaged across load. Results showed that MS RKN angle was significantly greater during WK (mean =174.63°, se =1.16°) and FM (mean =173.27°, se =2.16°) when compared to RN (mean =158.24°, se =1.80°) locomotion trial, averaged across levels of load($p<.001$).

There was a significant interaction between the effect of load and locomotion on MS RANK angle ($F(4, 44)= 3.20$, $p=.022$, $\eta^2_p =.225$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during RN ($F(1.34, 14,73) =3.72$, $p =.040$, $\eta_p^2 =2.53$). In order to find the pattern of differences on MS RANK angle depending on load during RN, post hoc marginal pairwise comparisons were performed. Results revealed that MS RANK angle was significantly greater at BW (mean =107.01°, se =1.65°) when compared to +25%BW (mean =106.00, se =1.79°) and +45%BW (mean =104.92°, se =1.63°) load conditions during RN ($p\leq.039$). There was also a significant main effect of load during FM ($F(2, 22) =6.08$, $p =.008$, $\eta_p^2 =.356$) Pairwise comparisons showed that MS RANK angle was significantly greater at BW (mean =113.17°, se =1.43°) when compared to +45%BW (mean =110.36°, se =1.57°) load condition during FM ($p=.005$). Additionally, simple main of locomotion on MS RANK angle were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22)=23.23$, $p<.001$, $\eta^2_p=.679$), +25%BW ($F(2, 22)=31.31$, $p<.001$, $\eta^2_p=.740$), +45%BW ($F(2, 22)=36.18$, $p<.001$, $\eta^2_p=.767$)]. To identify the pattern of differences on MS RANK angle depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on MS RANK angle were similar at each level of load. Results revealed that MS RANK angle was significantly greater during WK [BW (mean=112.16°, se=1.45°), +25%BW (mean=112.38°, se=1.36°), +45%BW (mean=110.79°, se=1.34°)] and FM [BW (mean=113.17°, se=1.43°), +25%BW (mean=112.06°, se=1.48°), +45%BW (mean=110.36°, se=1.57°)]when compared to RN [BW (mean=107.01°, se=1.65°), +25%BW (mean=106.00°, se=1.79°), +45%BW (mean=104.92°, se=1.63°)] at each level of load ($p<.001$).

4.4.3 Toe Off

There was no significant interaction between load and locomotion, in their effect on TO RHP angle ($F(4, 44) =.866$, $p =.492$, $\eta_p^2 =.073$).There was no significant main effect of load on TO RHP angle, averaged across levels of locomotion ($F(2, 22) =.106$, $p =.900$, $\eta_p^2 =.010$). However, there was a significant main effect of locomotion on TO RHP angle, averaged across levels of load ($F(2, 22)= 16.33$, $p<.001$, $\eta^2_p =.597$). In order to find the pattern of differences on TO RHP angle depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that TO RHP angle was significantly greater during WK (mean =182.31°, se =2.13°) when compared to both RN (mean =176.89°, se =2.46°) and FM (mean =175.98°, se =2.33°) locomotion trials, averaged across levels of load ($p\leq.004$).

There was a significant interaction between the effect of load and locomotion on TO RKN angle ($F(4, 44)= 4.89$, $p=.002$, $\eta^2_p =.308$). Simple main effects of load were analyzed for each level of locomotion. There was a significant main effect of load during WK ($F(2, 22)=26.65$, $p<.001$, $\eta^2_p=.708$). To identify the pattern of differences on TO RKN angle depending on load during WK, post hoc marginal pairwise comparisons were performed. Results revealed that TO RKN angle was significantly different for all load conditions, with BW (mean=148.22°, se=1.38°) eliciting the greatest TO RKN angle and +45%BW (mean=143.96°, se=1.64°) the lowest during WK ($p \leq .004$). Similar results were observed in the effects of load during RN ($F(2, 22)=42.75$, $p<.001$, $\eta^2_p=.795$). Pairwise comparisons revealed that TO RKN angle was significantly different for all load conditions, with BW (mean=146.28°, se=2.00°) eliciting the greatest TO RKN angle and +45%BW (mean=138.89°, se=1.96°) the lowest during RM ($p<.001$). There was also a significant main effect of load during FM ($F(2, 22)=19.86$, $p<.001$, $\eta^2_p=.644$). Pairwise comparisons showed that TO RKN angle was significantly greater at BW (mean=148.18°, se=1.77°) when compared to +25%BW (mean=144.27°, se=1.80°) and +45%BW (mean=142.91°, se=1.60°) load conditions during RM ($p \leq .001$). Additionally, simple main effects of locomotion on TO RKN angle were analyzed for each level of load. There was a significant main effect of locomotion with +25%BW ($F(2, 22)=6.09$, $p=.008$, $\eta^2_p=.356$). To identify the pattern of differences on TO RKN angle depending on locomotion with +25%BW, post hoc marginal pairwise comparisons were performed. Results showed that TO RKN angle was significantly greater during WK (mean=146.50°, se=1.52°) when compared to RN (mean= 141.78°, se=2.14°) with +25%BW ($p=.003$). There was also a significant main effect of locomotion with +45%BW ($F(2, 22)=9.48$, $p=.001$, $\eta^2_p=.463$). Pairwise comparisons revealed that TO RKN angle was significantly greater during both WK (mean=143.96°, se=1.64°) and FM (mean=142.91°, se=1.60°) when compared to RN (mean=138.89°, se=1.96°) with +45%BW ($p \leq .019$).

There was a significant interaction between the effect of load and locomotion on TO RANK angle ($F(4, 44)= 12.48$, $p<.001$, $\eta^2_p =.531$). Simple main effects of load were analyzed at each level of locomotion. There was no significant main effect of load at any level of locomotion.

Additionally, simple main effects of locomotion on TO RANK angle were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(2, 22)=85.91$, $p<.001$, $\eta^2_p=.886$). To identify the pattern of differences on TO RANK angle depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results showed that TO RANK angle was significantly different for all locomotion trials, with WK (mean=121.39°, se=1.22°) eliciting the greatest TO RANK angle and RN (mean=112.27°, se=1.52°) the lowest at BW ($p\leq.022$). There was also a significant main effect of locomotion with +25%BW ($F(2, 22)=165.16$, $p<.001$, $\eta^2_p=.938$). Pairwise comparisons showed that TO RANK angle was significantly greater during both WK (mean=121.20°, se=1.27°) and FM (mean=120.45°, se=1.43°) when compared to RN (mean=113.54°, se=1.38°) with +25%BW ($p<.001$). Lastly, there was a significant main effect of locomotion with +45%BW ($F(2, 22)=89.21$, $p<.001$, $\eta^2_p=.890$). Pairwise comparisons revealed that TO RANK angle was significantly different for all locomotion trials, with WK (mean=120.19°, se=1.40°) eliciting the greatest TO RANK angle and RN (mean=114.48°, se=1.57°) the lowest with +45%BW ($p\leq.021$).

Table 12. Descriptive statistics for right lower limb joint angles in degrees at heel strike, mid stance and toe off across all load and locomotion conditions

HIP

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	168.04	10.84	168.41	159.60 - 174.00
	25%BW	12	167.78	10.90	169.28	161.08 - 176.80
	45%BW	12	167.22	11.83	166.05	158.13 - 176.31
RN	BW	12	166.97	11.96	164.75	158.89 - 171.27
	25%BW	12	165.37	12.25	162.62	156.72 - 173.26
	45%BW	12	165.82	12.25	164.16	158.29 - 169.66
FM $^{g, gg}$	BW	12	162.07	12.20	158.92	155.54 - 169.99
	25%BW	12	162.42	13.09	161.72	151.37 - 172.21
	45%BW	12	161.50	11.82	158.94	153.83 - 166.90

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	201.64	9.09	202.49	192.83 - 206.23
	25%BW	12	202.39	6.69	200.23	198.20 - 209.63
	45%BW	12	202.19	8.63	204.18	195.35 - 206.81
RN ^{g}	BW	12	191.15	9.75	189.40	182.94 - 198.14
	25%BW	12	191.65	8.21	190.53	185.18 - 196.68
	45%BW	12	194.59	8.84	193.76	188.42 - 198.72
FM $^{g, gg}$	BW	12	198.41	7.48	198.95	190.37 - 204.49
	25%BW	12	197.85	8.93	199.16	191.42 - 205.17
	45%BW	12	199.32	7.71	199.01	194.22 - 201.58

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	182.35	8.26	182.52	177.61 - 185.80
	25%BW	12	182.72	7.38	181.92	175.80 - 189.29
	45%BW	12	181.86	9.65	177.83	174.58 - 189.99
RN ^{g}	BW	12	177.46	10.15	176.02	170.68 - 183.13
	25%BW	12	176.39	8.91	175.86	169.30 - 181.45
	45%BW	12	176.81	9.15	176.24	171.00 - 180.59
FM ^{g}	BW	12	176.91	9.15	174.92	170.06 - 184.25
	25%BW	12	175.70	9.06	173.43	167.87 - 183.81
	45%BW	12	175.33	9.45	173.10	167.94 - 182.37

KNEE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	163.93	6.57	163.40	160.95 - 167.24
	25%BW	12	160.88 ^{w}	8.01	161.28	154.63 - 163.47
	45%BW	12	159.18 ^{w}	8.36	160.25	153.94 - 164.34
RN ^{g}	BW	12	150.03	7.24	150.07	143.84 - 156.04
	25%BW	12	146.52 ^{w}	7.28	146.21	141.49 - 153.67
	45%BW	12	145.34 ^{w}	7.28	144.80	141.42 - 151.13
FM $^{g, gg}$	BW	12	156.24	8.95	157.58	147.38 - 163.20
	25%BW	12	152.03 ^{w}	9.49	150.62	143.16 - 159.67
	45%BW	12	150.66 ^{w}	8.35	148.80	143.43 - 157.78

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	175.70	4.76	175.31	170.75 - 179.08
	25%BW	12	174.70	3.72	175.57	171.85 - 177.70
	45%BW	12	173.49 ^{w}	4.55	172.60	170.48 - 178.08
RN ^{g}	BW	12	159.18	6.98	160.11	153.50 - 164.64
	25%BW	12	157.98	6.54	158.00	153.89 - 163.94
	45%BW	12	157.56 ^{w}	5.74	156.61	154.16 - 161.52
FM ^{gg}	BW	12	174.84	7.63	174.98	169.91 - 181.77
	25%BW	12	173.19	7.85	175.36	167.35 - 178.45
	45%BW	12	171.79 ^{w}	7.68	171.36	168.99 - 176.02

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	148.22	4.78	149.46	144.19 - 152.08
	25%BW	12	146.50 ^{w}	5.28	146.82	141.84 - 150.74
	45%BW ^{gg}	12	143.96 ^{w,ww}	5.69	145.66	138.21 - 148.74
RN	BW	12	146.28	6.94	146.70	140.36 - 153.55
	25%BW ^{g}	12	141.78 ^{w}	7.41	142.40	135.41 - 147.54
	45%BW ^{g}	12	138.89 ^{w,ww}	6.78	139.85	134.25 - 143.97
FM	BW	12	148.18	6.13	150.25	142.48 - 153.21
	25%BW	12	144.27 ^{w}	6.23	143.35	137.95 - 150.70
	45%BW ^{gg}	12	142.91 ^{w}	5.53	142.31	137.14 - 148.11

ANKLE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	120.16	4.33	119.64	117.52 - 123.13
	25%BW	12	120.16	4.09	120.81	117.89 - 122.52
	45%BW	12	118.88	4.92	118.85	116.40 - 121.54
RN ^{g}	BW	12	108.12	5.61	108.94	102.57 - 114.06
	25%BW	12	109.40	5.09	109.34	105.64 - 113.97
	45%BW	12	110.24	5.41	109.61	107.36 - 113.54
FM $^{g, gg}$	BW	12	117.08	4.25	116.17	114.59 - 119.46
	25%BW	12	117.18	4.82	117.34	114.37 - 120.17
	45%BW	12	116.34	4.98	116.78	112.23 - 119.29

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	112.16	5.03	110.91	108.29 - 117.70
	25%BW	12	112.38	4.72	110.80	108.87 - 117.18
	45%BW	12	110.79	4.64	110.13	107.89 - 112.96
RN ^{g}	BW	12	107.01	5.71	105.89	101.63 - 113.38
	25%BW	12	106.00 ^{w}	6.18	105.19	101.13 - 112.08
	45%BW	12	104.92 ^{w}	5.65	105.11	100.47 - 108.63
FM ^{gg}	BW	12	113.17	4.96	113.04	109.25 - 116.23
	25%BW	12	112.06	5.13	111.37	107.03 - 117.09
	45%BW	12	110.36 ^{w}	5.43	110.22	105.58 - 113.75

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	121.39	4.22	120.29	117.31 - 125.81
	25%BW ^{gg}	12	121.20	4.41	121.40	118.07 - 124.53
	45%BW	12	120.19	4.83	121.59	116.07 - 122.71
RN	BW ^{g}	12	112.27	5.27	112.35	106.97 - 117.91
	25%BW ^{g}	12	113.54	4.78	113.80	109.08 - 117.92
	45%BW ^{g}	12	114.48	5.44	114.01	110.24 - 117.47
FM	BW $^{g, gg}$	12	120.08	4.38	119.48	116.35 - 122.32
	25%BW ^{gg}	12	120.45	4.96	121.75	116.04 - 124.35
	45%BW $^{g, gg}$	12	119.12	5.39	119.03	115.23 - 122.61

^{w} significantly different from BW ^{ww} significantly different from +25%BW ^{g} significantly different from WK ^{gg} significantly different from RN

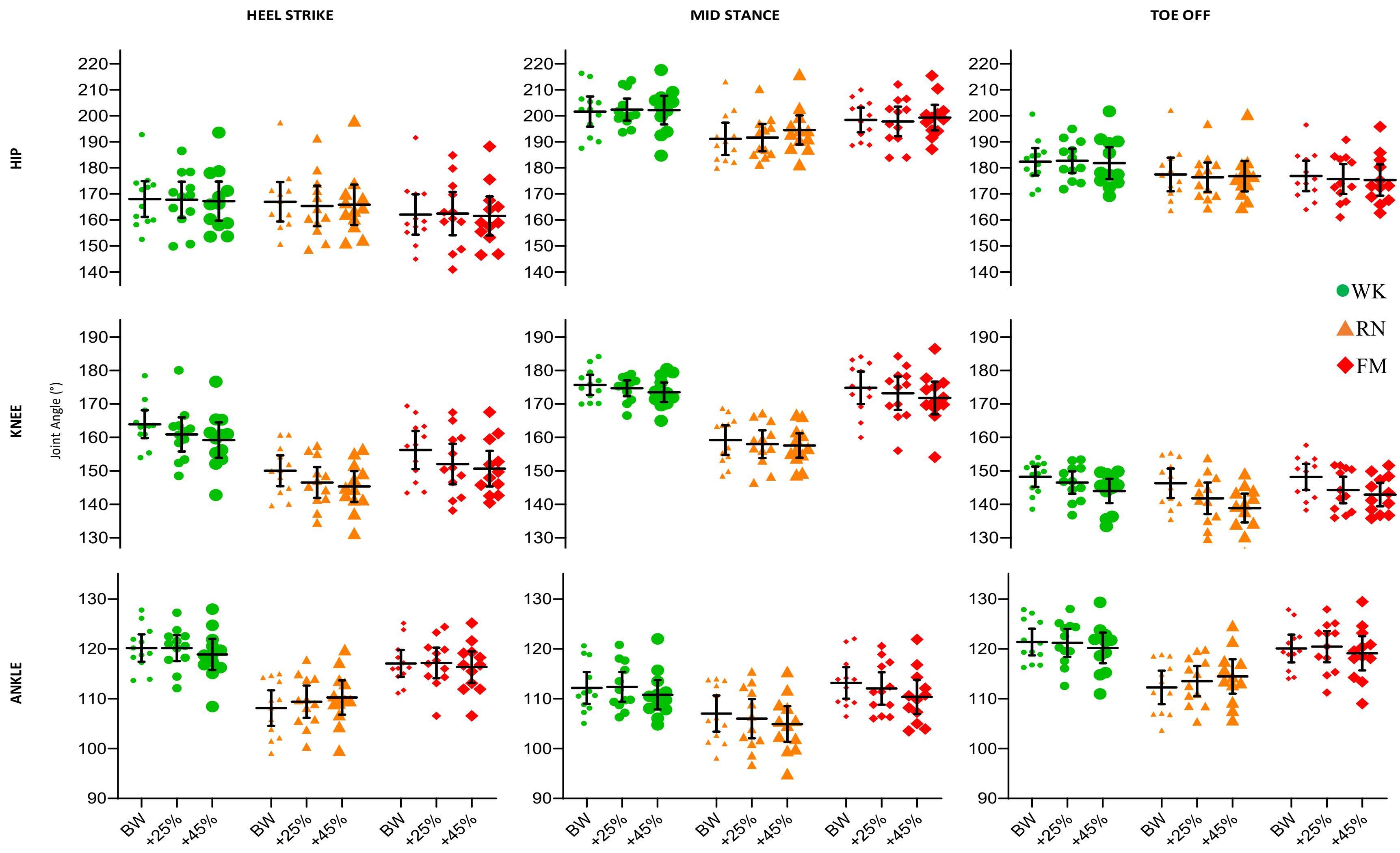

Figure 10. Right lower limb mean joint angles in degrees with 95% CI at heel strike, mid stance and toe off across all load and locomotion conditions

4.5 Left Leg Kinetics

3 x 3 two-way repeated measures ANOVA were conducted to examine the effects of load (BW, +25%BW, +45%BW) and locomotion (WK, RN, FM) on left hip (LHP), knee (LKN) and ankle (LANK) moments at heel strike (HS), mid stance (MS), and toe off (TO). Results reported below are grouped by gait event and ordered by joint from superior to inferior. Joint moment units are reported in newton meters per kilogram (Nm/kg). Values reported for joint moment follow the right-hand rule. For the hip, positive values represent a flexion moment and negative values represent an extension moment. At the knee, positive values represent an extension moment and negative values represent a flexion moment. Lastly, at the ankle, positive values represent a flexion moment (dorsiflexion) and negative values represent and extension moment (plantarflexion). **Table 12** and **Figure 9** in this section depict all the results for left leg kinetics.

4.5.1 Heel Strike

There was a significant interaction between the effects of load and locomotion on HS LHP moment ($F(2.18, 23.99) = 9.22$, $p < .001$, $\eta_p^2 = 456$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 12.81$, $p < .001$, $\eta_p^2 = .538$). In order to find the pattern of differences on HS LHP moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that HS LHP moment was significantly greater at BW (mean = -.707, se =.065 Nm/kg) and +25%BW (mean = -.647, se =.066 Nm/kg) than with +45%BW (mean = -.563, se =.060 Nm/kg) during WK

($p \leq .003$). Similar results were seen in the effect of load during FM ($F(2, 22) = 5.84$, $p = .009$, $\eta_p^2 = .347$). Pairwise comparisons revealed that HS LHP moment was significantly greater at BW (mean = -.986, se =.075 Nm/kg) and +25%BW (mean = -.961, se =.082 Nm/kg) than with +45%BW (mean = -.881, se =.082 Nm/kg) during FM ($p \leq .014$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(1.34, 14.75) = 30.27$, $p < .001$, $\eta_p^2 = .733$). In order to find the pattern of differences on HS LHP moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results showed that HS LHP moment was significantly greater at during FM (mean = -.986, se =.075 Nm/kg) than both WK (mean = -.707, se =.065 Nm/kg) and RN (mean = -.683, se =.048 Nm/kg) locomotion trials at BW ($p < .001$). There was also a significant main effect of locomotion with +25%BW ($F(2, 22) = 36.21$, $p < .001$, $\eta_p^2 = .767$). Pairwise comparisons showed that HS LHP moment was significantly different for all locomotion trials, with FM (mean = -.961, se =.082 Nm/kg) eliciting the greatest HS LHP moment and WK (mean = -.647, se =.066 Nm/kg) the lowest with +25%BW ($p \leq .029$). Similar results were seen in the effect of locomotion with +45%BW ($F(2, 22) = 41.83$, $p < .001$, $\eta_p^2 = .792$). Pairwise comparisons showed that HS LHP moment was significantly different for all locomotion trials, with FM (mean = -.881, se =.082 Nm/kg) eliciting the greatest HS LHP moment and WK (mean = -.563, se =.060 Nm/kg) the lowest with +45%BW ($p \leq .007$).

There was a significant interaction between the effects of load and locomotion on HS LKN moment ($F(4, 44) = 11.02$, $p < .001$, $\eta^2_p = .501$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 7.18$, $p = .004$, $\eta_p^2 = .395$). In order to find the pattern of differences on HS LKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that

HS LKN moment was significantly greater at BW (mean = -.428, se =.046 Nm/kg) and +25%BW (mean = -.403, se =.033 Nm/kg) than at the +45%BW (mean = -.354, se=.033 Nm/kg) load condition during WK ($p \leq .009$). There was also a significant main effect of load during RN ($F(2, 22) = 8.18$, $p = .002$, $\eta_p^2 = .426$). Pairwise comparisons revealed that HS LKN moment was significantly greater with +25%BW (mean = -.346, se =.027 Nm/kg) and +45%BW (mean = -.364, se =.027 Nm/kg) than at BW (mean = -.295, se =.030 Nm/kg) during RN ($p \leq .018$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(2, 22) = 22.64$, $p < .001$, $\eta_p^2 = .673$). In order to find the pattern of differences on HS LKN moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results showed that HS LKN moment was significantly greater during WK (mean = -.428, se =.046 Nm/kg) and FM (mean = -.495, se =.038 Nm/kg) than during RN (mean = -.295, se =0.30 Nm/kg) at BW ($p \leq .001$). There was also a significant main effect of locomotion with +25%BW ($F(2, 22) = 12.81$, $p < .001$, $\eta_p^2 = .538$). Pairwise comparisons revealed that HS LKN moment was significantly different for each locomotion trial, with FM (mean = -.462, se =.031 Nm/kg) eliciting the greatest HS LKN moment and RN (mean = -.346, se =.027 Nm/kg) the lowest with +25%BW ($p \leq .036$). Lastly, there was a significant main effect of locomotion with +45%BW ($F(2, 22) = 16.34$, $p < .001$, $\eta_p^2 = .598$). Pairwise comparisons showed that HS LKN moment was significantly greater during FM (mean = -.447, se =.029 Nm/kg) than both WK (mean = -.354, se =.033 Nm/kg) and RN (mean = -.364, se =.027 Nm/kg) locomotion trials with +45%BW ($p = .001$).

There was no significant interaction between load and locomotion, in their effects on HS LANK moment ($F(2.29, 25.23) = .387$, $p = .817$, $\eta_p^2 = .034$). Furthermore, there were no significant main effects of load, averaged across locomotion ($F(2, 22) = 2.66$, $p = .092$, $\eta_p^2 = .195$), or of locomotion, averaged across load ($F(2, 22) = 2.31$, $p = .123$, $\eta_p^2 = .174$) on HS LANK moment.

4.5.2 Mid Stance

There was a significant interaction between the effects of load and locomotion on MS LHP moment ($F(4, 44) = 6.34$, $p < .001$, $\eta^2_p = .366$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during RN ($F(2, 22) = 8.49$, $p = .002$, $\eta_p^2 = .436$). In order to find the pattern of differences on MS LHP moment depending on load during RN, post hoc marginal pairwise comparisons were performed. Results showed that MS LHP moment was significantly greater at BW (mean =.317, se =.070 Nm/kg) when compared to +25%BW (mean =.200, se =.053 Nm/kg) and +45%BW (mean =.147, se =.052 Nm/kg) load conditions during RN ($P \leq .043$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(2, 22) = 27.01$, $p < .001$, $\eta_p^2 = .711$). In order to find the pattern of differences on MS LHP moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results revealed that MS LHP moment was significantly greater during RN (mean =.317, se =.070 Nm/kg) when compared to WK (mean =.062, se =.063 Nm/kg) and FM (mean = -.017, se =.043 Nm/kg) locomotion trials at BW ($p < .001$). There was also a significant main effect of locomotion at +25%BW ($F(2, 22) = 17.66$, $p < .001$, $\eta_p^2 = .616$). Pairwise comparisons showed that MS LHP moment was significantly different for all locomotion trials, with RN (mean =.200, se =.053 Nm/kg) eliciting the greatest MS LHP moment and FM (mean = -.026, se =.051 Nm/kg) the lowest with +25%BW ($p \leq .03$). Similar results were seen in the effect of locomotion with +45%BW ($F(1.31, 14.45) = 13.56$, $p = .001$, $\eta_p^2 = .551$). Pairwise comparisons showed that MS LHP moment was significantly different for all locomotion trials, with RN (mean =.147, se =.053 Nm/kg) eliciting the greatest MS LHP moment and FM (mean = -.025, se =.029 Nm/kg) the lowest with +45%BW ($p \leq .46$).

There was a significant interaction between the effects of load and locomotion on MS LKN moment ($F(2.11, 23.17) = 15.64$, $p < .001$, $\eta^2_p = .587$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 3.82$, $p = .038$, $\eta_p^2 = .258$). In order to find the pattern of differences on MS LKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that MS LKN moment was significantly greater at BW (mean = -.187, se =.048 Nm/kg) and +25%BW (mean = -.165, se =.029 Nm/kg) than with the +45%BW (mean = -.115, se =.028 Nm/kg) load condition during WK ($p \leq .031$). There was also a significant main effect of load during RN ($F(2, 22) = 15.69$, $p < .001$, $\eta_p^2 = .588$). Pairwise comparisons revealed that MS LKN moment was significantly different for all load conditions, with BW (mean=1.417, se=.157 Nm/kg) eliciting the greatest MS LKN moment and +45%BW (mean =.911, se =.102 Nm/kg) the lowest during RN ($p \leq .009$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(1.27, 14.02) = 97.87$, $p < .001$, $\eta_p^2 = .899$), +25%BW ($F(1.16, 12.74) = 110.217$, $p < .001$, $\eta_p^2 = .909$), +45%BW ($F(2, 22) = 66.54$, $p < .001$, $\eta_p^2 = .858$)]. In order to find the pattern of differences on MS LKN moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effect of locomotion on MS LKN moment were similar at each level of load. Results showed that MS LKN moment was significantly greater during RN [BW (mean =1.417, se =.157 Nm/kg), +25%BW (mean =1.179, se =.119 Nm/kg), +45%BW (mean =.911, se =.102 Nm/kg)] when compared to WK [BW (mean = -.187, se =.048 Nm/kg), +25%BW (mean = -.165, se =.029 Nm/kg), +45%BW (mean = -.115, se =.028 Nm/kg)] and FM [BW (mean = -.190, se=.057 Nm/kg), +25%BW (mean = -.132, se=.037 Nm/kg), +45%BW (mean = -.102, se =.062 Nm/kg)] locomotion trials at each level of load ($p<.001$).

There was a significant interaction between the effects of load and locomotion on MS LANK moment ($F(4, 44) =51.39$, $p<.001$, $\eta^2_p =.824$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during RN ($F(2, 22) =95.84$, $p<.001$, $\eta_p^2 =.897$). In order to find the pattern of differences on MS LANK moment depending on load during RN, post hoc marginal pairwise comparisons were performed. Results revealed that MS LANK moment was significantly different for all load conditions, with BW (mean = -2.371, se =.143 Nm/kg) eliciting the greatest MS LANK moment and +45%BW (mean = -1.66, se =.130 Nm/kg) the lowest during RN ($p<.001$). Additionally, simple main effects of locomotion on MS LANK moment were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(1.29, 14.24) =118.29$, $p<.001$, $\eta_p^2 =.915$), +25%BW ($F(1.35, 14.81) =112.60$, $p<.001$, $\eta_p^2 =.911$), +45%BW ($F(1.21, 13.31) =51.09$, $p<.001$ $\eta_p^2 =.823$)]. In order to find the pattern of differences on MS LANK moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effect of locomotion on MS LANK moment were similar at each level of load. Results showed that MS LANK moment was significantly greater during RN [BW (mean = -2.371, se =.143 Nm/kg), +25%BW (mean = -1.916, se =.144 Nm/kg), +45%BW (mean = -1.661, se =.130 Nm/kg)] when compared to WK [BW (mean = -.576, se =.039 Nm/kg), +25%BW (mean = -.508, se =.027 Nm/kg), +45%BW (mean = -.527, se =.029 Nm/kg)] and FM [BW (mean = -.572, se =.077 Nm/kg), +25%BW (mean = -.603, se =.063 Nm/kg), +45%BW (mean = -.614, se =.075 Nm/kg)] locomotion trials at each level of load ($p<.001$).

4.5.3 Toe Off

There was no significant interaction between load and locomotion, in their effects on TO LHP moment ($F(1.77, 19.45) = 1.16$, $p = .329$, $\eta_p^2 = .095$). There was a significant main effect of load on TO LHP moment, averaged across levels of locomotion ($F(2, 22) = 6.52$, $p = .006$, $\eta^2_p = .372$). In order to find the pattern of differences on TO LHP moment depending on load, averaged across locomotion, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results showed that TO LHP moment was significantly greater at BW (mean = -.198, se =.030 Nm/kg) than with +45%BW (mean = -.145, se =.033 Nm/kg) load, averaged across levels of locomotion ($p = .01$). Additionally, there was a significant main effect of locomotion on TO LHP moment, averaged across levels of load ($F(2, 22) = 47.16$, $p < .001$, $\eta^2_p = .811$). In order to find the pattern of differences on TO LHP moment depending on locomotion, averaged across load, post hoc marginal pairwise comparisons were performed using the Bonferroni adjustment. Results revealed that TO LHP moment was significantly greater during both RN (mean = -263, se =.027 Nm/kg) and FM (mean = -.219, se =.045 Nm/kg) than during WK (mean = -.041, se =.026 Nm/kg), averaged across levels of load ($p < .001$).

There was a significant interaction between the effects of load and locomotion on TO LKN moment ($F(4, 44) = 4.01$, $p = .007$, $\eta^2_p = .267$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 9.31$, $p = .001$, $\eta_p^2 = .458$). In order to find the pattern of differences on TO LKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that TO LKN moment was significantly greater at BW (mean = -.150, se =.019 Nm/kg) and +25%BW (mean = -.137, se =.013 Nm/kg) when compared to +45%BW (mean = -.111, se =.014 Nm/kg) load condition during WK ($p \leq .02$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22) = 3.62$, $p=.044$, $\eta_p^2 = .248$), +25%BW ($F(2, 22) = 9.69$, $p=.001$, $\eta_p^2 = .468$), +45%BW ($F(2, 22) = 30.08$, $p<.001$, $\eta_p^2 = .732$)]. In order to find the pattern of differences on TO LKN moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on TO LKN moment were similar at each level of load. Results revealed that TO LKN moment was significantly greater during RN [BW (mean = -.185, se =.018 Nm/kg), +25%BW (mean = -.204, se =.018 Nm/kg), +45%BW (mean = -.198, se =.018 Nm/kg)] and FM [BW (mean = -.198, se =.019 Nm/kg), +25%BW (mean = -.178, se =.018 Nm/kg), +45%BW (mean = -.169, se =.015 Nm/kg)] when compared to WK [BW (mean = -.150, se =.019 Nm/kg), +25%BW (mean = -.137, se=.013 Nm/kg), +45%BW (mean = -.111, se =.014 Nm/kg)] at each level of load ($p \leq .32$).

There was no significant interaction between load and locomotion, in their effects on TO LANK moment ($F(2.26, 24,87) = .31$, $p=.765$, $\eta_p^2 = .027$). Furthermore, there were no significant main effects of load, averaged across locomotion ($F(1.35, 14.80) = 1.65$, $p=.223$, $\eta_p^2 = .131$), or of locomotion, averaged across load ($F(2, 22) = 2.75$, $p=.086$, $\eta_p^2 = .200$) on TO LANK moment.

Table 13. Descriptive statistics for left lower limb joint moment in newton meters per kilogram at heel strike, mid stance and toe off across all load and locomotion conditions

HIP

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.71	0.23	-0.78	-0.91 - -0.48
	25%BW	12	-0.65	0.23	-0.70	-0.77 - -0.59
	45%BW	12	-0.56 ^{w,ww}	0.21	-0.61	-0.74 - -0.46
RN	BW	12	-0.68	0.17	-0.68	-0.85 - -0.51
	25%BW^{g}	12	-0.74	0.18	-0.77	-0.83 - -0.59
	45%BW^{g}	12	-0.74	0.16	-0.78	-0.87 - -0.59
FM	BW$^{g, gg}$	12	-0.99	0.26	-1.05	-1.18 - -0.74
	25%BW$^{g, gg}$	12	-0.96	0.28	-1.00	-1.19 - -0.78
	45%BW$^{g, gg}$	12	-0.88 ^{w,ww}	0.29	-0.89	-1.12 - -0.78

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW^{gg}	12	0.06	0.22	0.05	-0.08 - 0.18
	25%BW	12	0.05	0.17	-0.01	-0.04 - 0.13
	45%BW	12	0.06	0.13	0.04	-0.02 - 0.09
RN	BW^{g}	12	0.32	0.24	0.31	0.10 - 0.49
	25%BW^{g}	12	0.20 ^{w}	0.18	0.20	0.05 - 0.28
	45%BW^{g}	12	0.15 ^{w}	0.18	0.19	-0.02 - 0.29
FM	BW^{gg}	12	-0.02	0.15	-0.02	-0.09 - 0.03
	25%BW$^{g, gg}$	12	-0.03	0.18	-0.06	-0.17 - 0.12
	45%BW$^{g, gg}$	12	-0.03	0.10	-0.05	-0.09 - 0.02

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.06	0.09	-0.08	-0.13 - 0.00
	25%BW	12	-0.04	0.09	-0.06	-0.07 - -0.04
	45%BW	12	-0.02 ^{w}	0.09	-0.04	-0.09 - 0.03
RN ^{g}	BW	12	-0.30	0.09	-0.29	-0.38 - -0.23
	25%BW	12	-0.26	0.10	-0.26	-0.29 - -0.18
	45%BW	12	-0.23 ^{w}	0.11	-0.26	-0.30 - -0.10
FM ^{g}	BW	12	-0.23	0.16	-0.25	-0.30 - -0.14
	25%BW	12	-0.23	0.16	-0.25	-0.34 - -0.16
	45%BW	12	-0.19 ^{w}	0.16	-0.23	-0.34 - -0.08

KNEE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW^{gg}	12	-0.43	0.16	-0.47	-0.55 - -0.26
	25%BW	12	-0.40	0.11	-0.42	-0.49 - -0.37
	45%BW	12	-0.35 ^{w,ww}	0.11	-0.39	-0.43 - -0.26
RN	BW^{g}	12	-0.29	0.11	-0.28	-0.35 - -0.23
	25%BW^{g}	12	-0.35 ^{w}	0.09	-0.35	-0.41 - -0.25
	45%BW	12	-0.36 ^{w}	0.09	-0.38	-0.43 - -0.28
FM	BW^{gg}	12	-0.49	0.13	-0.50	-0.60 - -0.37
	25%BW$^{g, gg}$	12	-0.46	0.11	-0.45	-0.55 - -0.37
	45%BW$^{g, gg}$	12	-0.45	0.10	-0.43	-0.51 - -0.41

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	-0.19	0.17	-0.19	-0.34 - -0.09
	25%BW	12	-0.17	0.10	-0.15	-0.26 - -0.10
	45%BW	12	-0.12 $^{w, ww}$	0.10	-0.12	-0.21 - -0.03
RN ^{g}	BW	12	1.42	0.54	1.36	1.06 - 1.98
	25%BW	12	1.18 ^{w}	0.41	1.14	0.85 - 1.47
	45%BW	12	0.91 $^{w, ww}$	0.35	0.83	0.61 - 1.28
FM ^{gg}	BW	12	-0.19	0.20	-0.22	-0.29 - -0.12
	25%BW	12	-0.13	0.13	-0.15	-0.22 - -0.03
	45%BW	12	-0.10 ^{w}	0.21	-0.16	-0.24 - -0.04

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.15	0.06	-0.17	-0.20 - -0.10
	25%BW	12	-0.14	0.05	-0.15	-0.17 - -0.13
	45%BW^{gg}	12	-0.11 ^{w,ww}	0.05	-0.13	-0.16 - -0.06
RN ^{g}	BW	12	-0.19	0.06	-0.19	-0.23 - -0.14
	25%BW^{g}	12	-0.20 ^{w}	0.06	-0.20	-0.25 - -0.15
	45%BW^{g}	12	-0.20 ^{w,ww}	0.06	-0.20	-0.26 - -0.14
FM ^{g}	BW	12	-0.20	0.07	-0.22	-0.25 - -0.14
	25%BW	12	-0.18 ^{w}	0.06	-0.16	-0.24 - -0.13
	45%BW^{gg}	12	-0.17 ^{w}	0.05	-0.16	-0.22 - -0.12

ANKLE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.01	0.11	-0.05	-0.08 - 0.08
	25%BW	12	0.00	0.09	-0.02	-0.06 - 0.05
	45%BW	12	0.03	0.07	0.05	-0.03 - 0.08
RN	BW	12	-0.01	0.16	0.04	-0.05 - 0.08
	25%BW	12	0.01	0.11	0.06	-0.04 - 0.08
	45%BW	12	0.01	0.10	0.06	-0.05 - 0.07
FM	BW	12	0.07	0.16	0.02	-0.05 - 0.20
	25%BW	12	0.10	0.10	0.08	0.03 - 0.21
	45%BW	12	0.09	0.10	0.07	0.00 - 0.19

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	-0.58	0.14	-0.57	-0.72 - -0.44
	25%BW	12	-0.51	0.09	-0.48	-0.61 - -0.44
	45%BW	12	-0.53	0.10	-0.54	-0.61 - -0.44
RN ^{g}	BW	12	-2.37	0.50	-2.41	-2.68 - -2.04
	25%BW	12	-1.92 ^{w}	0.39	-1.83	-2.26 - -1.69
	45%BW	12	-1.66 $^{w, ww}$	0.45	-1.55	-1.87 - -1.38
FM ^{gg}	BW	12	-0.57	0.27	-0.58	-0.70 - -0.36
	25%BW	12	-0.60	0.22	-0.59	-0.73 - -0.40
	45%BW	12	-0.61	0.26	-0.58	-0.69 - -0.46

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	0.02	0.04	0.02	-0.01 - 0.05
	25%BW	12	0.03	0.04	0.02	0.01 - 0.05
	45%BW	12	0.04	0.03	0.04	0.04 - 0.06
RN	BW	12	0.01	0.12	0.05	0.02 - 0.07
	25%BW	12	0.02	0.08	0.05	0.01 - 0.06
	45%BW	12	0.02	0.07	0.05	-0.01 - 0.06
FM	BW	12	0.07	0.10	0.03	-0.01 - 0.18
	25%BW	12	0.09	0.06	0.10	0.04 - 0.15
	45%BW	12	0.08	0.06	0.09	0.03 - 0.14

^{w} significantly different from BW ^{ww} significantly different from +25%BW ^{g} significantly different from WK ^{gg} significantly different from RN

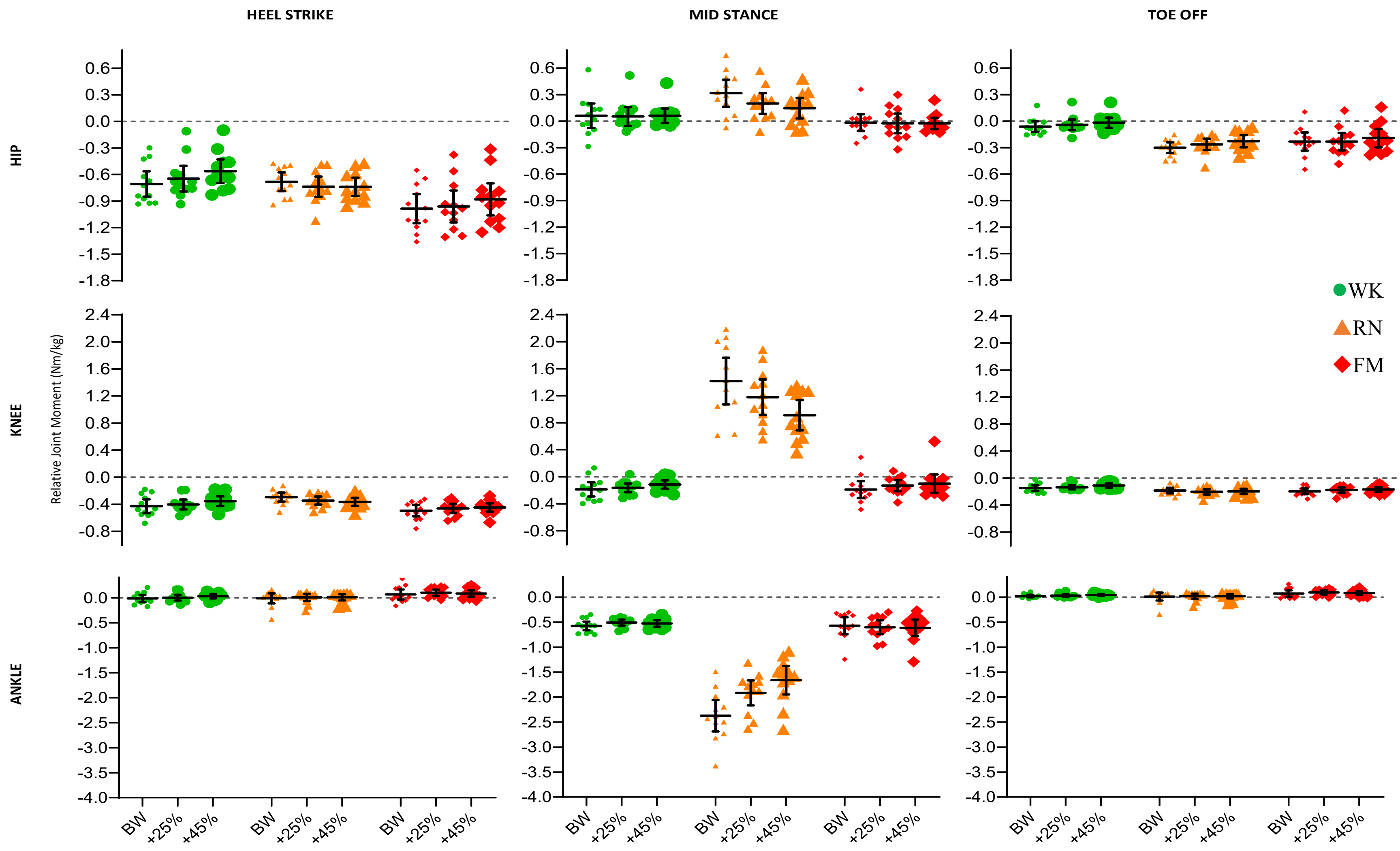

Figure 11. Left lower limb mean joint moment in newton meters per kilogram with 95% CI at heel strike, mid stance and toe off across all load and locomotion conditions

4.6 Right Leg Kinetics

3 x 3 two-way repeated measures ANOVA were conducted to examine the effects of load (BW, +25%BW, +45%BW) and locomotion (WK, RN, FM) on right hip (RHP), knee (RKN) and ankle (RANK) angles at heel strike (HS), mid stance (MS), and toe off (TO). Results reported below are grouped by gait event and ordered by joint from superior to inferior. Joint moment units are reported in newton meters per kilogram (Nm/kg). Values reported for joint moment follow the right-hand rule. For the hip, positive values represent a flexion moment and negative values represent an extension moment. At the knee, positive values represent an extension moment and negative values represent a flexion moment. Lastly, at the ankle, positive values represent a flexion moment (dorsiflexion) and negative values represent and extension moment (plantarflexion). **Table 13** and **Figure 10** in this section depict all the results for right leg kinematics.

4.6.1 Heel Strike

There was a significant interaction between the effects of load and locomotion on HS RHP moment ($F(4, 44) = 17.44$, $p < .001$, $\eta^2_p = .613$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 8.81$, $p = .002$, $\eta_p^2 = .445$). In order to find the pattern of differences on HS RHP moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that HS RHP moment was significantly greater at BW (mean = -1.047, se =.097 Nm/kg) and +25%BW (mean = -.980, se =.103 Nm/kg) than with +45%BW (mean = -.878, se =.095 Nm/kg) during WK

($p \leq .043$). Similar results were seen in the effect of load during FM ($F(2, 22) = 14.43$, $p < .001$, $\eta_p^2 = .567$). Pairwise comparisons revealed that HS RHP moment was significantly greater at BW (mean = -1.378, se =.091 Nm/kg) and +25%BW (mean = -1.329, se =.101 Nm/kg) than with +45%BW (mean = -1.208, se =.096 Nm/kg) during FM ($p \leq .007$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(2, 22) = 36.84$, $p < .001$, $\eta_p^2 = .770$). In order to find the pattern of differences on HS RHP moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results showed that HS RHP moment was significantly different for all locomotion trials, with FM (mean = -1.378, se =.091 Nm/kg) eliciting the greatest HS RHP moment and RN (mean = -.748, se =.049 Nm/kg) the lowest at BW ($p \leq .004$). Similar results were seen in the effect of locomotion with +25%BW ($F(1.18, 12.97) = 33.47$, $p < .001$, $\eta_p^2 = .753$). Pairwise comparisons showed that HS RHP moment was significantly different for all locomotion trials, with FM (mean = -1.329, se =.101 Nm/kg) eliciting the greatest HS RHP moment and RN (mean = -.787, se =.050 Nm/kg) the lowest with +25%BW ($p < .05$). Lastly, there was a significant main effect of locomotion with +45%BW ($F(1.32, 15.54) = 19.82$, $p < .001$, $\eta_p^2 = .643$). Pairwise comparisons showed that HS RHP moment was significantly greater during FM (mean = -1.208, se =.096 Nm/kg) than during WK (mean = -.878, se =.095 Nm/kg) and RN (mean = -.814, se =.040 Nm/kg) locomotion trials with +45%BW ($p \leq .001$).

There was a significant interaction between the effects of load and locomotion on HS RKN moment ($F(4, 44) = 16.37$, $p < .001$, $\eta^2_p = .598$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 3.84$, $p = .037$, $\eta_p^2 = .259$). In order to find the pattern of differences on HS RKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that

HS RKN moment was significantly greater at BW (mean = -.597, se =.061 Nm/kg) and +25%BW (mean = -.579, se =.056 Nm/kg) than at the +45%BW (mean = -.531, se =.051 Nm/kg) load condition during WK ($p \leq .04$). There was also a significant main effect of load during RN ($F(2, 22) = 25.68$, $p < .001$, $\eta_p^2 = .700$). Pairwise comparisons revealed that HS RKN moment was significantly different at each level of locomotion, with +45%BW (mean = -.424, se =.017 Nm/kg) eliciting the greatest HS RKN moment and BW (mean = -.315, se =.028 Nm/kg) the lowest during RN ($p \leq .049$). Lastly, there was a significant main effect of load during FM ($F(2, 22) = 5.04$, $p = .016$, $\eta_p^2 = .314$). Pairwise comparisons showed that HS RKN moment was significantly greater at BW (mean = -.715, se =.053 Nm/kg) and +25%BW (mean = -.700, se =.042 Nm/kg) than at the +45%BW (mean = -.644, se =.044 Nm/kg) load condition during FM ($p \leq .026$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22) = 35.18$, $p < .001$, $\eta_p^2 = .762$), +25%BW ($F(2, 22) = 25.96$, $p < .001$, $\eta_p^2 = .702$), +45%BW ($F(2, 22) = 12.93$, $p < .001$, $\eta_p^2 = .540$)]. In order to find the pattern of differences on HS RKN moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on HS RKN moment were similar at each level of load. Results revealed that HS RKN moment was significantly different for each locomotion trial, with FM [BW (mean = -.715, se =.053 Nm/kg), +25%BW (mean = -.700, se =.042 Nm/kg), +45%BW (mean = -.644, se =.044 Nm/kg)] eliciting the greatest HS RKN moment and RN [BW (mean = -.315, se =.028 Nm/kg), +25%BW (mean = -.385, se =.027 Nm/kg), +45%BW (mean = -.424, se =.017 Nm/kg)] the lowest at each level of load ($p \leq .005$)

There was no significant interaction between load and locomotion, in their effects on HS RANK moment ($F(2.18, 24.01) = .30$, $p = .877$, $\eta_p^2 = .026$). Furthermore, there were no significant main effects of load, averaged across locomotion ($F(2, 22) = .44$, $p = .648$, $\eta_p^2 = .039$), or of locomotion, averaged across load ($F(2, 22) = .701$, $p = .507$, $\eta_p^2 = .060$) on HS RANK moment.

4.6.2 Mid Stance

There was a significant interaction between the effects of load and locomotion on MS RHP moment ($F(4, 44) = 3.71$, $p = .011$, $\eta^2_p = .252$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during RN ($F(2, 22) = 6.43$, $p = .006$, $\eta_p^2 = .369$). In order to find the pattern of differences on MS RHP moment depending on load during RN, post hoc marginal pairwise comparisons were performed. Results showed that MS RHP moment was significantly greater at BW (mean =.135, se =.112 Nm/kg) when compared to +45%BW (mean = -.017, se =.087 Nm/kg) load condition during RN ($P = .001$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(1.11, 12.24) = 4.66$, $p = .020$, $\eta_p^2 = .298$). In order to find the pattern of differences on MS RHP moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results revealed that MS RHP moment was a significantly greater extension moment during FM (mean = -.123, se =.053 Nm/kg) when compared to WK (mean = -.033, se =.053 Nm/kg) and RN (mean =.135, se =.112 Nm/kg) locomotion trials at BW ($p \leq .04$). There was also a significant main effect of locomotion at +25%BW ($F(2, 22) = 5.79$, $p = .010$, $\eta_p^2 = .345$). Pairwise comparisons showed that MS RHP moment was significantly greater in extension during FM (mean = -.159, se =.058 Nm/kg) when compared to both WK (mean = -.056, se =.045 Nm/kg) and RN (mean =.054, se =.086 Nm/kg) locomotion trials with +25%BW ($p \leq .032$).

There was a significant interaction between the effects of load and locomotion on MS RKN moment ($F(2.07, 22.74) = 22.10$, $p<.001$, $\eta^2_p = .668$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 5.85$, $p=.009$, $\eta_p^2 = .346$). In order to find the pattern of differences on MS RKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that MS RKN moment was significantly greater at BW (mean = -.238, se =.040 Nm/kg) than with the +45%BW (mean = -.164, se =.033 Nm/kg) load condition during WK ($p=.006$). There was also a significant main effect of load during RN ($F(2, 22) = 21.96$, $p<.001$, $\eta_p^2 = .666$). Pairwise comparisons revealed that MS RKN moment was significantly different for all load conditions, with BW (mean =1.31, se =.176 Nm/kg) eliciting the greatest MS RKN moment and +45%BW (mean =.752, se =.100 Nm/kg) the lowest during RN ($p \leq .007$). Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22) = 76.28$, $p<.001$, $\eta_p^2 = .874$), +25%BW ($F(1.19, 13.10) = 73.83$, $p<.001$, $\eta_p^2 = .870$), +45%BW ($F(2, 22) = 58.11$, $p<.001$, $\eta_p^2 = .841$)]. In order to find the pattern of differences on MS RKN moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effects of locomotion on MS RKN moment were similar at each level of load. Results showed that MS RKN moment was significantly greater in extension during RN [BW (mean =1.308, se =.176 Nm/kg), +25%BW (mean =1.049, se =.145 Nm/kg), +45%BW (mean =.752, se =.100 Nm/kg)] when compared to WK [BW (mean = -.238, se =.040 Nm/kg), +25%BW (mean = -.210, se =.032 Nm/kg), +45%BW (mean = -.164, se =.033 Nm/kg)] and FM [BW (mean = -.219, se =.069 Nm/kg), +25%BW (mean = -.184, se =.029 Nm/kg), +45%BW (mean = -.119, se =.068)] locomotion trials at each level of load ($p<.001$).

There was a significant interaction between the effects of load and locomotion on MS RANK moment ($F(4, 44) = 38.44$, $p<.001$, $\eta^2_p = .778$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(1.26, 13.90) = 5.38$, $p=.013$, $\eta_p^2 = .328$). In order to find the pattern of differences on MS RANK moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that MS RANK moment was significantly greater at BW (mean = -.650, se =.024 Nm/kg) than with both +25%BW (mean = -.581, se =.028 Nm/kg) and +45%BW (mean = -.596, se =.026 Nm/kg) load conditions during WK ($p \leq .006$). There was also a significant main effect of load during RN ($F(2, 22) = 64.11$, $p<.001$, $\eta_p^2 = .854$). Pairwise comparisons revealed that MS RANK moment was significantly different for all load conditions, with BW (mean = -2.482, se =.178 Nm/kg) eliciting the greatest MS RANK moment and +45%BW (mean = -1.767, se =.146 Nm/kg) the lowest during RN ($p<.001$). Additionally, simple main effects of locomotion on MS RANK moment were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(1.25, 13.76) = 81.94$, $p<.001$, $\eta_p^2 = .882$), +25%BW ($F(2, 22) = 93.22$, $p<.001$, $\eta_p^2 = .894$), +45%BW ($F(1.19, 13.04) = 43.73$, $p<.001$, $\eta_p^2 = .799$)]. In order to find the pattern of differences on MS RANK moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effect of locomotion on MS RANK moment were similar at each level of load. Results showed that MS RANK moment was significantly greater during RN [BW (mean = -2.482, se =.178 Nm/kg), +25%BW (mean = -2.077, se =.123 Nm/kg), +45%BW (mean = -1.767, se =.146 Nm/kg)] when compared to WK [BW (mean = -.650, se =.024 Nm/kg), +25%BW (mean = -.581, se =.028 Nm/kg), +45%BW (mean = -.596, se =.026 Nm/kg)] and FM [BW (mean = -.666, se =.091 Nm/kg), +25%BW (mean = -.663, se=.079

Nm/kg), +45%BW (mean = -.679, se =.078 Nm/kg)] locomotion trials at each level of load ($p<.001$).

4.6.3 Toe Off

There was a significant interaction between load and locomotion, in their effects on TO RHP moment ($F(4, 44) = 2.84$, $p=.035$, $\eta^2_p = .205$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 8.36$, $p=.002$, $\eta_p^2 = .432$). In order to find the pattern of differences on TO RHP moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that TO RHP moment was significantly greater at BW (mean = -.284, se =.041 Nm/kg) when compared to +45%BW (mean = -.202, se =.038 Nm/kg) load condition during WK ($p=.001$). There was also a significant main effect of load during RN ($F(2, 22) = 6.00$, $p=.008$, $\eta_p^2 = .353$). Pairwise comparisons revealed that TO RHP moment was significantly greater at BW (mean = -.331, se =.025 Nm/kg) than both +25%BW (mean = -.283, se =.028 Nm/kg) and +45%BW (mean = -.273, se =.022 Nm/kg) load conditions during RN ($p\leq.016$). Lastly, there was a significant main effect of load during FM ($F(2, 22) = 14.02$, $p<.001$, $\eta_p^2 = .560$). Pairwise comparisons showed that TO RHP moment was significantly greater at both BW (mean = -.498, se =.047 Nm/kg) and +25%BW (mean = -.474, se =.055 Nm/kg) when compared to +45%BW (mean = -.409, se =.051 Nm/kg) load condition during FM ($p\leq.006$). Additionally, simple main effects of locomotion on TO RHP moment were analyzed at each level of load. There was a significant main effect of locomotion at each level of load [BW ($F(2, 22) = 23.30$, $p<.001$, $\eta_p^2 = .679$), +25%BW ($F(1.11, 12.25) = 29.38$, $p<.001$, $\eta_p^2 = .728$), +45%BW ($F(1.25, 13.71) = 21.55$, $p<.001$, $\eta_p^2 = .662$)]. In order to find the pattern of differences on TO RHP moment depending on locomotion at each level of load, post hoc marginal pairwise comparisons were performed. The effect of locomotion on TO RHP moment were similar at each level of load. Results showed that TO RHP moment was significantly greater during FM [BW (mean = -.498, se =.047 Nm/kg), +25%BW (mean = -.474, se =.055 Nm/kg), +45%BW (mean = -.409, se =.051 Nm/kg)] when compared to WK [BW (mean = -.284, se =.041 Nm/kg), +25%BW (mean = -.249, se =.041 Nm/kg), +45%BW (mean = -.202, se =.038 Nm/kg)] and RN [BW (mean = -.331, se =.025 Nm/kg), +25%BW (mean = -.283, se=.028 Nm/kg), +45%BW (mean = -.273, se =.022 Nm/kg)] locomotion trials at each level of load ($p \leq .009$).

There was a significant interaction between the effects of load and locomotion on TO RKN moment ($F(4, 44) = 12.34$, $p < .001$, $\eta^2_p = .529$). Simple main effects of load were analyzed at each level of locomotion. There was a significant main effect of load during WK ($F(2, 22) = 5.02$, $p = .016$, $\eta_p^2 = .313$). In order to find the pattern of differences on TO RKN moment depending on load during WK, post hoc marginal pairwise comparisons were performed. Results showed that TO RKN moment was significantly greater at BW (mean = -.271, se =.028 Nm/kg) and +25%BW (mean = -.258, se =.023 Nm/kg) when compared to +45BW (mean = -.230, se =.022 Nm/kg) load condition during WK ($p \leq .048$). There was also a significant main effect of load during RN $F(2, 22) = 11.45$, $p < .001$, $\eta_p^2 = .510$). Pairwise comparisons showed that TO RKN moment was significantly greater with +25%BW (mean = -.224, se =.017 Nm/kg) and +45%BW (mean = -.238, se =.011 Nm/kg) than at BW (mean = -.195, se =.016) during RN ($p \leq .002$). Lastly, there was a significant main effect of load during FM $F(2, 22) = 6.85$, $p = .005$, $\eta_p^2 = .384$). Pairwise comparisons revealed that TO RKN moment was significantly greater at BW (mean = -.363, se =.028 Nm/kg) and +25%BW (mean = -.343, se =.024 Nm/kg) when compared to +45%BW (mean = -.308, se=.023 Nm/kg) during FM. Additionally, simple main effects of locomotion were analyzed at each level of load. There was a significant main effect of locomotion at BW ($F(2, 22) = 22.52$,

$p<.001$, $\eta_p^2 = .672$). In order to find the pattern of differences on TO RKN moment depending on locomotion at BW, post hoc marginal pairwise comparisons were performed. Results revealed that TO RKN moment was significantly different for all locomotion trials, with FM (mean = -.363, se =.028 Nm/kg) eliciting the greatest TO RKN moment and RN (mean = -.195, se =.016 Nm/kg) the lowest at BW ($p \leq .01$). There was also a significant main effect of locomotion with +25%BW ($F(2, 22) = 14.16$, $p<.001$, $\eta_p^2 = .563$). Pairwise comparisons showed that TO RKN moment was significantly greater during FM (mean = -.343, se = .024 Nm/kg) when compared to both WK (mean = -.258, se =.023 Nm/kg) and RN (mean = -.224, se =.017 Nm/kg) locomotion trials with +25%BW ($p \leq .001$). Similar results were seen in the effect of locomotion with +45%BW ($F(2, 22) = 8.87$, $p=.002$, $\eta_p^2 = .446$). Pairwise comparisons showed that TO RKN moment was significantly greater during FM (mean = -.308, se =.023 Nm/kg) when compare to both WK (mean = -.230, se =.022 Nm/kg) and RN (mean = -.238, se =.011 Nm/kg) locomotion trials with +25%BW ($p \leq .015$).

There was no significant interaction between load and locomotion, in their effects on TO RANK moment ($F(2.12, 23.34) = .70$ $p=.596$, $\eta_p^2 = .060$). Furthermore, there were no significant main effects of load, averaged across locomotion ($F(2, 22) = .795$, $p=464$, $\eta_p^2 = .067$), or of locomotion, averaged across load ($F(1.32, 14.49) = .66$, $p=.529$, $\eta_p^2 = .056$) on TO RANK moment.

Table 14. Descriptive statistics for right lower limb joint moment in newton meters per kilogram at heel strike, mid stance and toe off across all load and locomotion conditions

HIP

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.03	0.09	-0.03	-0.10 - 0.03
	25%BW	12	-0.02	0.09	-0.03	-0.10 - 0.01
	45%BW	12	-0.01 ^{w,ww}	0.07	-0.04	-0.05 - 0.04
RN	BW^{g}	12	-0.01	0.17	0.05	0.00 - 0.08
	25%BW^{g}	12	-0.01	0.12	0.03	0.00 - 0.05
	45%BW	12	-0.01	0.12	0.02	-0.01 - 0.03
FM	BW$^{g, gg}$	12	0.01	0.12	-0.03	-0.09 - 0.14
	25%BW$^{g, gg}$	12	0.04	0.10	0.05	-0.02 - 0.12
	45%BW$^{g, gg}$	12	0.03 ^{w,ww}	0.08	0.06	-0.05 - 0.08

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.03	0.18	-0.03	-0.17 - 0.01
	25%BW	12	-0.06	0.16	-0.08	-0.12 - -0.02
	45%BW	12	-0.05	0.14	-0.08	-0.15 - 0.03
RN	BW	12	0.14	0.39	0.09	-0.09 - 0.50
	25%BW	12	0.05	0.30	-0.01	-0.22 - 0.30
	45%BW	12	-0.02 ^{w}	0.30	-0.04	-0.26 - 0.24
FM	BW$^{g, gg}$	12	-0.12	0.18	-0.11	-0.26 - -0.02
	25%BW$^{g, gg}$	12	-0.16	0.20	-0.11	-0.23 - -0.04
	45%BW	12	-0.12	0.14	-0.13	-0.19 - -0.03

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.28	0.14	-0.31	-0.38 - -0.17
	25%BW	12	-0.25	0.14	-0.26	-0.35 - -0.18
	45%BW	12	-0.20 ^{w}	0.13	-0.19	-0.33 - -0.11
RN	BW	12	-0.33	0.09	-0.30	-0.42 - -0.27
	25%BW	12	-0.28 ^{w}	0.10	-0.27	-0.32 - -0.23
	45%BW	12	-0.27 ^{w}	0.07	-0.27	-0.36 - -0.21
FM $^{g, gg}$	BW	12	-0.50	0.16	-0.55	-0.60 - -0.36
	25%BW	12	-0.47 ^{w}	0.19	-0.52	-0.57 - -0.38
	45%BW	12	-0.41 ^{w}	0.18	-0.45	-0.55 - -0.30

KNEE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.60	0.21	-0.58	-0.78 - -0.44
	25%BW	12	-0.58	0.19	-0.63	-0.74 - -0.46
	45%BW	12	-0.53 ^{w,ww}	0.18	-0.56	-0.69 - -0.41
RN ^{g}	BW	12	-0.31	0.10	-0.30	-0.42 - -0.26
	25%BW	12	-0.39 ^{w}	0.09	-0.38	-0.46 - -0.32
	45%BW	12	-0.42 ^{w,ww}	0.06	-0.42	-0.47 - -0.39
FM $^{g, gg}$	BW	12	-0.72	0.18	-0.72	-0.90 - -0.55
	25%BW	12	-0.70	0.14	-0.72	-0.80 - -0.60
	45%BW	12	-0.64 ^{w,ww}	0.15	-0.67	-0.76 - -0.50

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	-0.24	0.14	-0.23	-0.36 - -0.13
	25%BW	12	-0.21	0.11	-0.20	-0.32 - -0.13
	45%BW	12	-0.16 $^{w, ww}$	0.11	-0.18	-0.25 - -0.08
RN ^{g}	BW	12	1.31	0.61	1.31	0.83 - 1.72
	25%BW	12	1.05 ^{w}	0.50	0.95	0.65 - 1.29
	45%BW	12	0.75 $^{w, ww}$	0.35	0.70	0.52 - 1.01
FM ^{gg}	BW	12	-0.22	0.24	-0.23	-0.33 - -0.21
	25%BW	12	-0.18	0.10	-0.20	-0.24 - -0.13
	45%BW	12	-0.12	0.24	-0.17	-0.24 - -0.11

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.27	0.10	-0.25	-0.33 - -0.21
	25%BW	12	-0.26	0.08	-0.27	-0.32 - -0.21
	45%BW^{gg}	12	-0.23 ^{w,ww}	0.08	-0.23	-0.29 - -0.17
RN ^{g}	BW^{g}	12	-0.20	0.05	-0.18	-0.26 - -0.17
	25%BW	12	-0.22 ^{w}	0.06	-0.21	-0.27 - -0.19
	45%BW	12	-0.24 ^{w}	0.04	-0.24	-0.27 - -0.21
FM ^{g}	BW$^{g, gg}$	12	-0.36	0.10	-0.38	-0.44 - -0.27
	25%BW$^{g, gg}$	12	-0.34	0.08	-0.34	-0.43 - -0.28
	45%BW$^{g, gg}$	12	-0.31 ^{w,ww}	0.08	-0.33	-0.37 - -0.22

ANKLE

HEEL STRIKE

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.03	0.09	-0.03	-0.10 - 0.03
	25%BW	12	-0.02	0.09	-0.03	-0.10 - 0.01
	45%BW	12	-0.01	0.07	-0.04	-0.05 - 0.04
RN	BW	12	-0.01	0.17	0.05	0.00 - 0.08
	25%BW	12	-0.01	0.12	0.03	0.00 - 0.05
	45%BW	12	-0.01	0.12	0.02	-0.01 - 0.03
FM	BW	12	0.01	0.12	-0.03	-0.09 - 0.14
	25%BW	12	0.04	0.10	0.05	-0.02 - 0.12
	45%BW	12	0.03	0.08	0.06	-0.05 - 0.08

MID STANCE

Locomotion	Load	N	Mean	SD	Med	IQR
WK ^{gg}	BW	12	-0.65	0.08	-0.68	-0.71 - -0.57
	25%BW	12	-0.58 ^{w}	0.10	-0.58	-0.67 - -0.51
	45%BW	12	-0.60 ^{w}	0.09	-0.62	-0.66 - -0.51
RN ^{g}	BW	12	-2.48	0.62	-2.36	-2.73 - -2.17
	25%BW	12	-2.08 ^{w}	0.43	-2.08	-2.16 - -1.83
	45%BW	12	-1.77 $^{w, ww}$	0.51	-1.68	-1.88 - -1.49
FM ^{gg}	BW	12	-0.67	0.32	-0.63	-0.71 - -0.46
	25%BW	12	-0.66	0.27	-0.64	-0.78 - -0.46
	45%BW	12	-0.68	0.27	-0.66	-0.70 - -0.53

TOE OFF

Locomotion	Load	N	Mean	SD	Med	IQR
WK	BW	12	-0.02	0.04	-0.03	-0.05 - 0.02
	25%BW	12	-0.01	0.04	-0.03	-0.04 - 0.00
	45%BW	12	0.00	0.03	0.00	-0.03 - 0.01
RN	BW	12	0.01	0.14	0.05	0.02 - 0.07
	25%BW	12	0.01	0.09	0.03	0.02 - 0.05
	45%BW	12	0.00	0.09	0.02	0.01 - 0.04
FM	BW	12	0.01	0.08	-0.02	-0.06 - 0.10
	25%BW	12	0.03	0.06	0.04	0.00 - 0.08
	45%BW	12	0.03	0.05	0.03	-0.01 - 0.06

^{w} significantly different from BW ^{ww} significantly different from +25%BW ^{g} significantly different from WK ^{gg} significantly different from RN

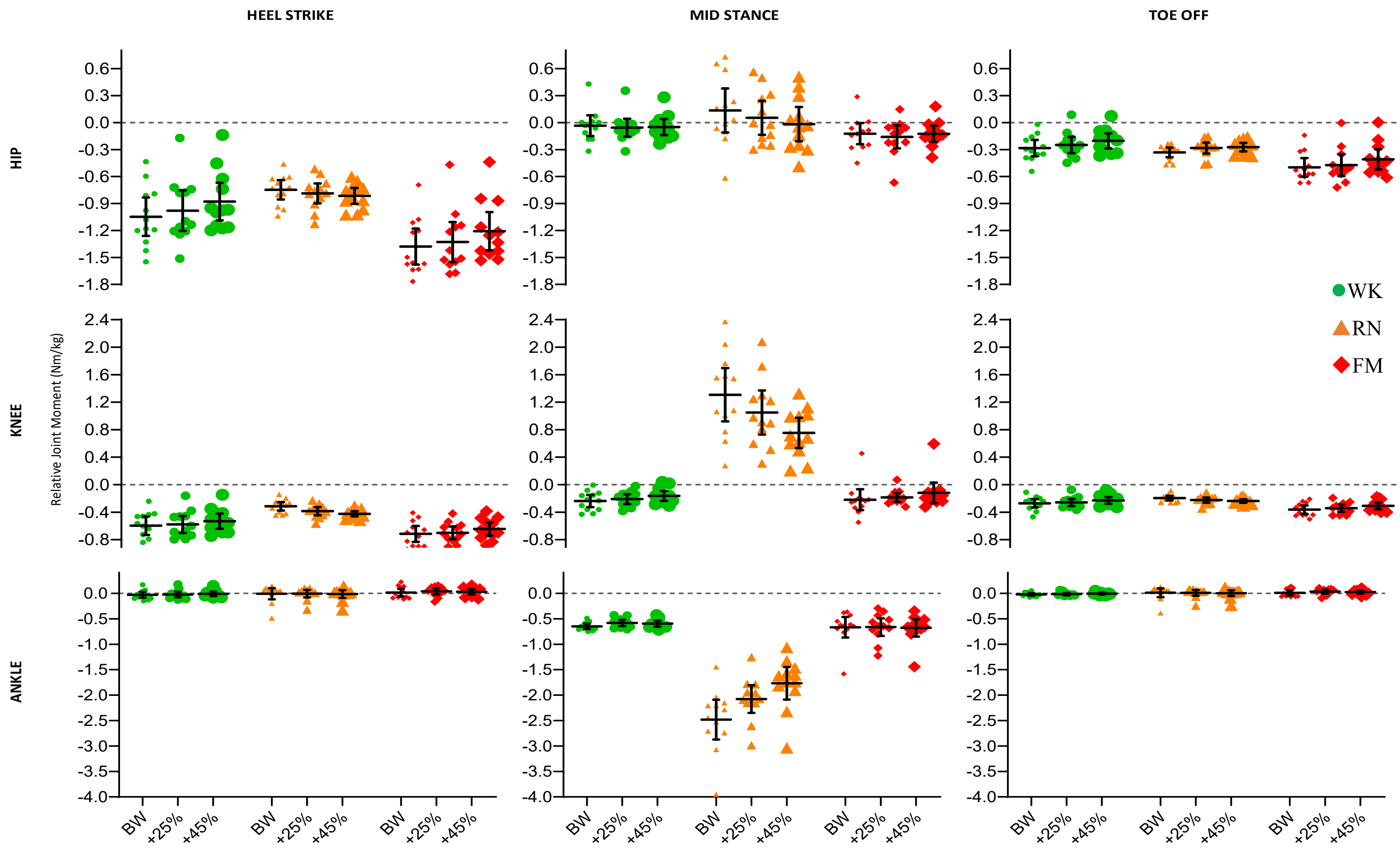

Figure 12. Right lower limb mean joint moment in newton meters per kilogram with 95% CI at heel strike, mid stance and toe off across all load and locomotion conditions

4.7 Coordination

No statistical analyses were conducted on coordination. Mean CRP was calculated for sagittal plane, shank-thigh coupling at each load condition for seven complete gait cycles. Standard deviations of the mean CRP were plotted to evaluate the variability of shank-thigh coupling throughout the gait cycle with increases in load magnitude across all levels of locomotion. Interpretation of CRP for the purposes of the current study is entirely observational. Two exemplar subjects were selected for CRP analysis. One who experienced the overall highest relative joint moments across all levels of load and locomotion and another who experienced the lowest. Figures 13 & 14 depict the CRP for shank-thigh coupling of left and right leg for Subject 04, who experienced the overall lowest relative joint moments. Increases in load do not appear to decrease the variability of shank-thigh coupling for Subject 04. Additionally, there is no apparent difference in the variability of shank-thigh coupling between RN and FM conditions. However, there appears to be less variability in shank-thigh coupling of the right leg compared to the left. Figures 15 & 16 depict the CRP for shank-thigh coupling of left and right leg for Subject 06, who experienced the overall highest relative joint moments. Increases in load appear to decrease variability of shank-thigh coupling for the left leg, however, this trend does not appear to hold true for the right leg. There is no discernable difference in the variability of shank-thigh coupling between RN and FM conditions. Similar to subject 06, there appears to be less variability in shank-thigh coupling of the right leg compared to the left. Overall Subject 04 appears to have greater shank-thigh coupling variability than subject 06

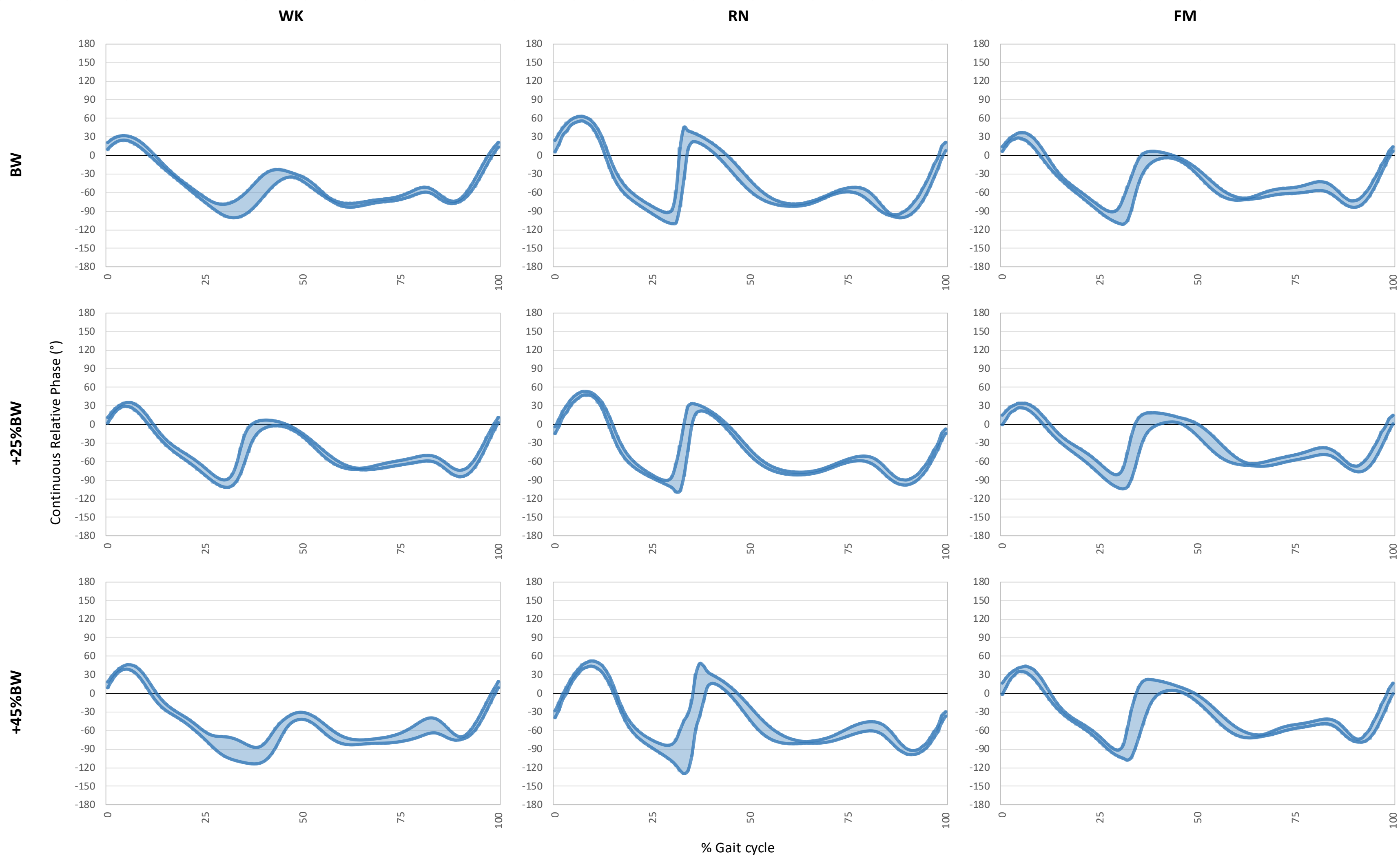

Figure 13. Subject 04 CRP for shank-thigh coupling of left leg at each load condition across all levels of locomotion.

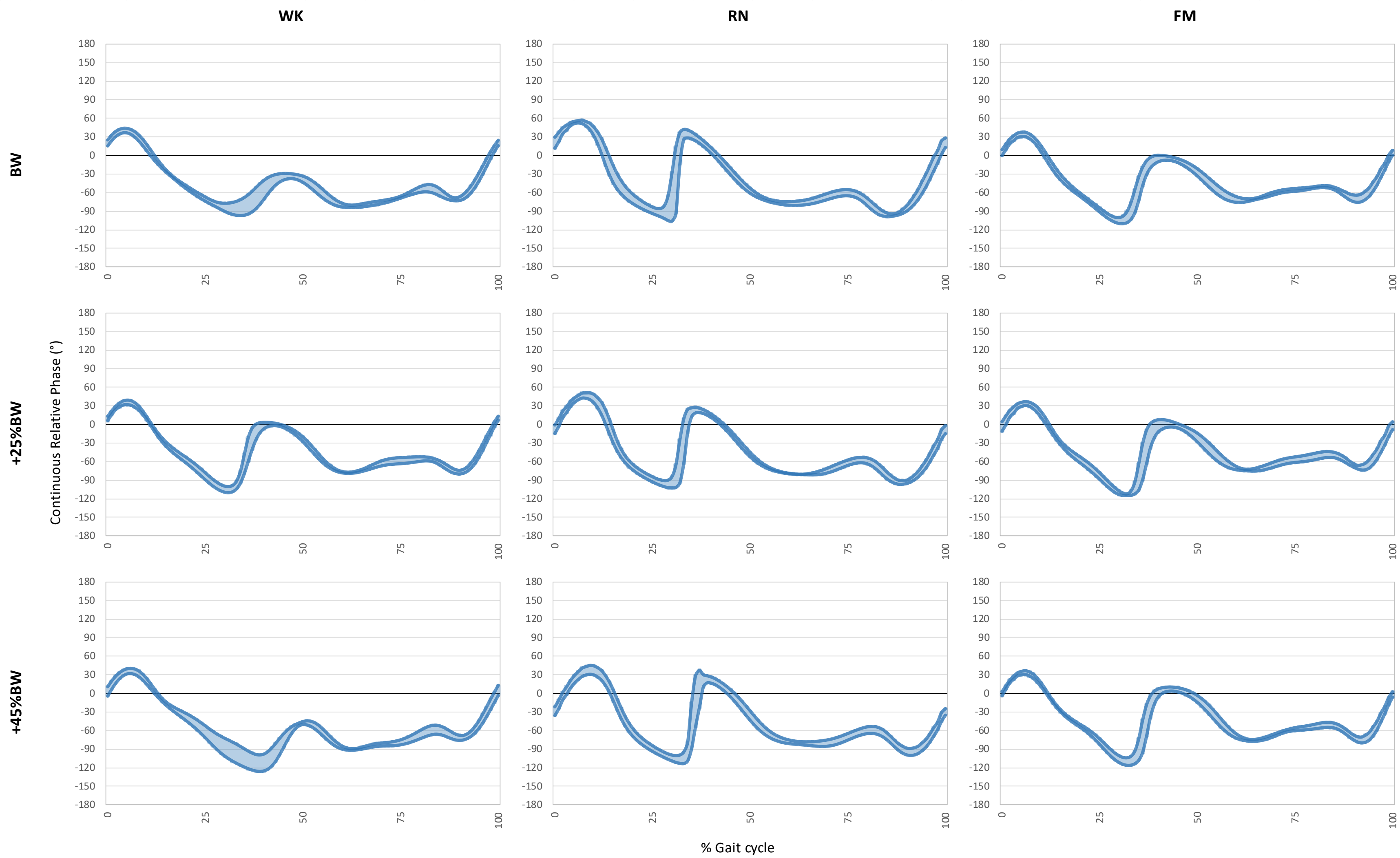

Figure 14. Subject 04 CRP for shank-thigh coupling of right leg at each load condition across all levels of locomotion.

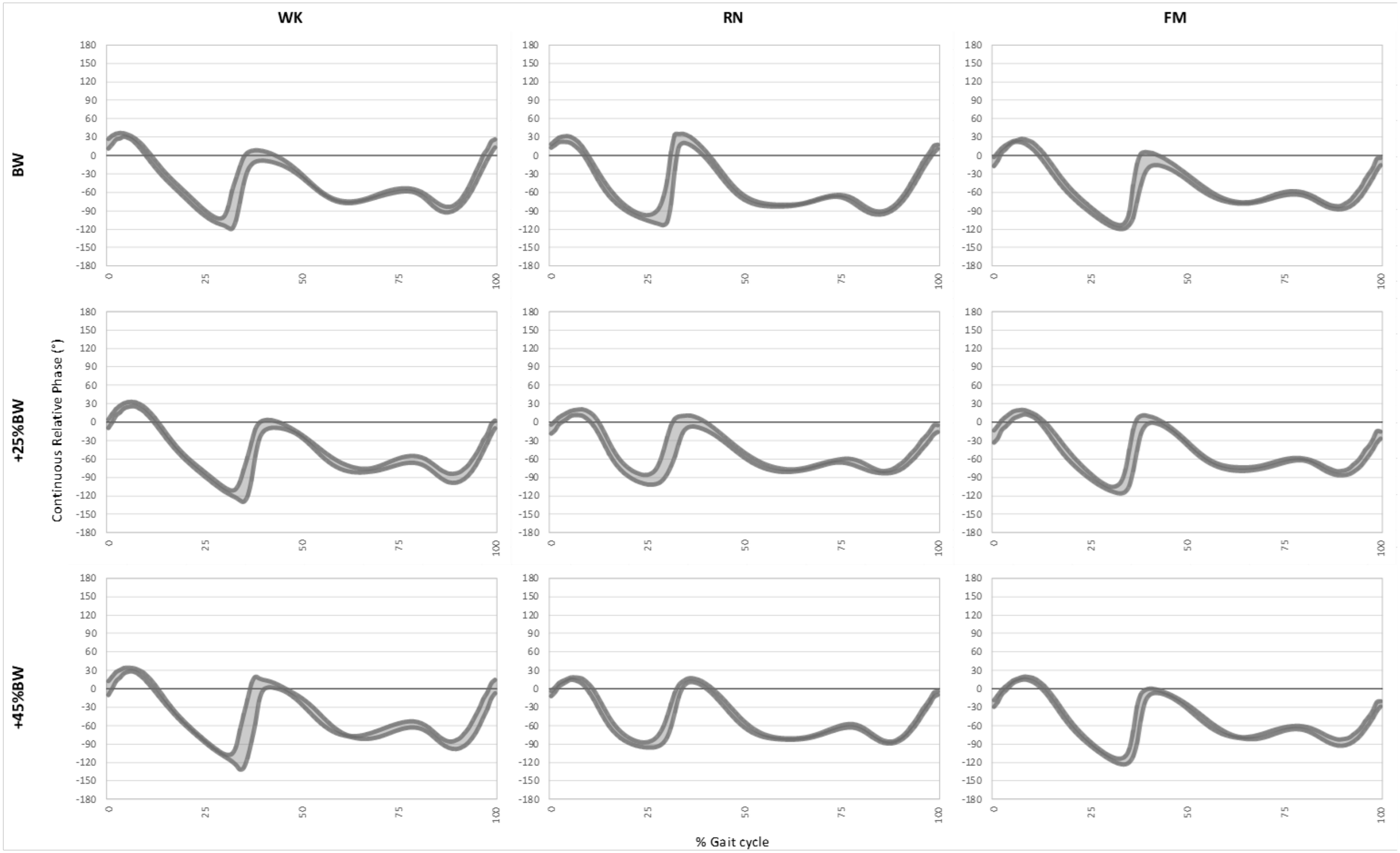

Figure 15. Subject 06 CRP for shank-thigh coupling of left leg at each load condition across all levels of locomotion.

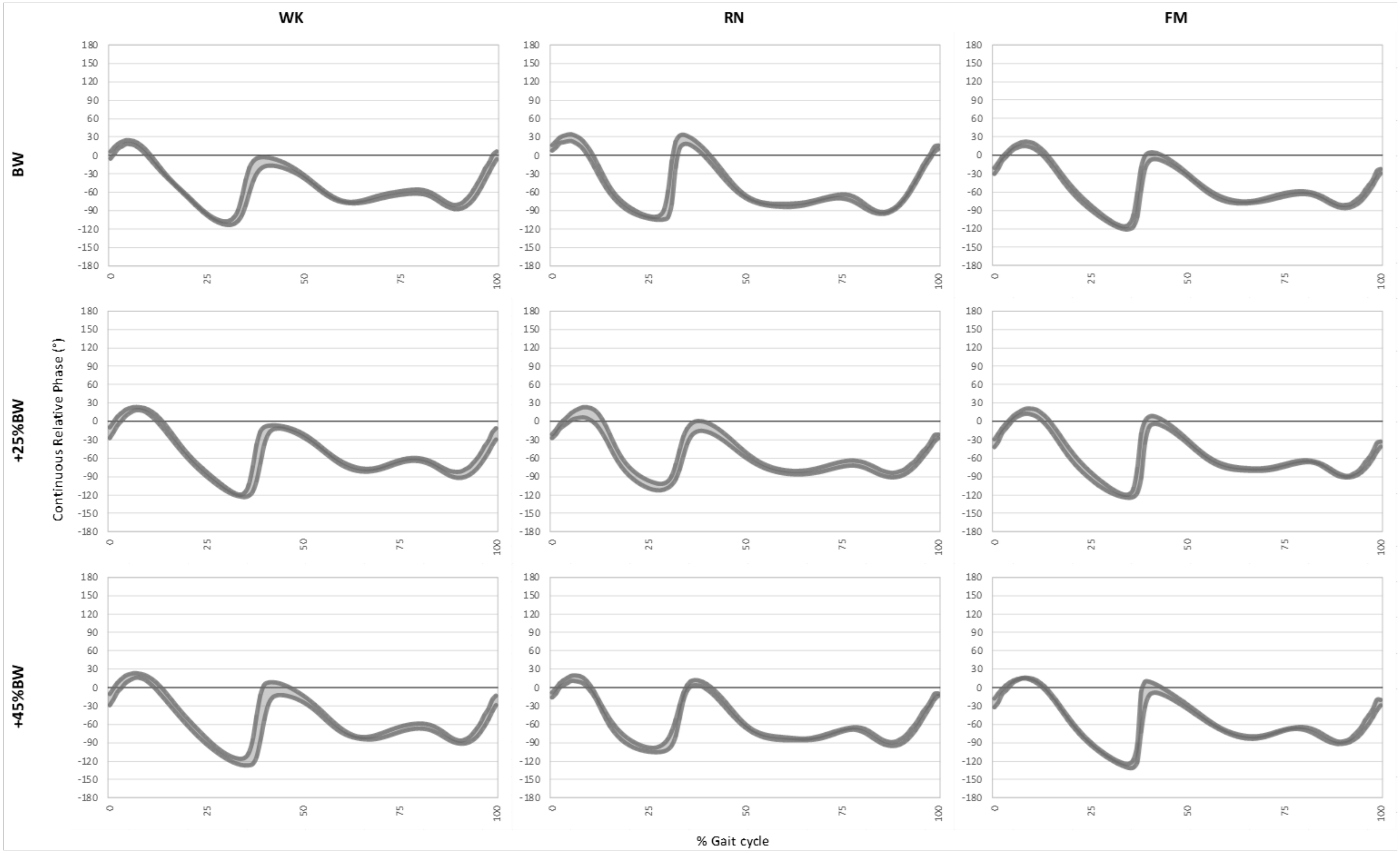

Figure 16. Subject 06 CRP for shank-thigh coupling of right leg at each load condition across all levels of locomotion.

5.0 Discussion

This is one of the first studies to investigate the interactive effects of combat load carriage magnitude on lower limb biomechanics during walking, running, and forced-marching in female subjects. Load carriage has been shown to alter gait biomechanics and place additional stress on the musculoskeletal system, augmenting the warfighters risk of sustaining a musculoskeletal injury. With the technological advancements and growing complexity of modern warfare, warfighters are continuously burdened with more gear. These heavier loads will conceivably induce greater musculoskeletal stress and magnify the warfighters injury risk. With female applicants now being considered for combat arms roles, there is a demand for studies investigating the effects of combat load carriage on the biomechanics of the female warfighter.

The primary purpose of this study was to investigate the main effects of load carriage mass on kinematic, kinetic, and spatiotemporal gait characteristics at marching velocities around the GTP in female subjects. As a secondary aim, coordination patterns were examined for each load and locomotion condition to determine levels of variability within the system. The data from a total of 12 subjects was used for analysis. All kinematic and kinetic outcomes were analyzed and reported by joint (hip, knee, ankle) at heel strike, mid stance, and toe off. Spatiotemporal metrics, kinematics, kinetics, and study limitations and future research are discussed with more detail in the sections below.

5.1 Spatiotemporal Metrics

The literature shows that, in general, increases in load carriage decrease stride length and flight time and increase stride width, stride frequency, stance time, and double support time. We hypothesized that our results would align with previous findings. Our results for spatial and temporal metrics were in nearly in complete agreement with previous research.(E. Harman et al., 2000; E. A. Harman et al., 2000; J. Knapik et al., 2010; Liew et al., 2016; Majumdar et al., 2010) Load carriage had a significant effect on stride length and width. With heavier loads, subjects effectively shorted their stride length and increased their stride width. This was likely an effort to increase stability with the added load. Shortening the stride length decreases the displacement of the body's CoM in the sagittal plane and increasing stride width creates a wider base of support, all of which increases stability.(C. T. Farley & D. P. Ferris, 1998) Locomotion also had a significant effect on stride length and width. Stride length was higher during FM when compared to WK and RN and stride width was higher for both WK and FM when compared to RN. These findings come as no surprise. Since the RN and FM conditions were conducted at the same treadmill velocity, 10% above the subject's GTP. Running at this velocity is more akin to a slow jog and would not demand a long stride. Additionally, a more flexed knee at heel strike is common during running which may contribute to the shorter stride length observed.(Mann & Hagy, 1980) It makes sense that maintaining a walk at these velocities would demand a significantly greater stride length. Increased hip adduction resulted in decreased stride width during running. Hip adduction is a common adaptation during running and has been associated with iliotibial band syndrome in females.(Foch & Milner, 2014) In the sagittal plane, the feet are more in line with the body's CoM throughout stance phase.(C. T. Farley & D. P. Ferris, 1998) This allows the propulsive forces generated from mid-stance to toe off to be applied more sagittally, driving the body forward while decreasing the displacement of the body's CoM in the frontal plane.(C. T. Farley & D. P. Ferris, 1998) No changes in stride frequency were observed with the addition of load. These findings do not align with previous research.(E. Harman et al., 2000; E. A. Harman et al., 2001; E. A. Harman et al., 2000; J. Knapik et al., 2010; Martin & Nelson, 1986) Martin and Nelson found that women consistently shortened their stride length and increased stride frequency with increases in load.(Martin & Nelson, 1986) Harman et al. observed similar results in their study on female soldiers. However, both previously mentioned studies controlled for marching velocity across load conditions. The velocities for each load condition in the current study were calculated from percentages of the subject's GTP for that load condition. Results for GTP velocity showed that subjects transitioned to running at a significantly lower velocity with +45%BW (1.75 m/s) than they did at BW (1.86 m/s) and +25%BW (1.83 m/s). Since the majority of subjects transitioned to running earlier with added load, the prescribed velocities across load conditions were slightly different. This may explain why load magnitude had no effect on stride frequency in our study. Had the velocities at each load condition been constant, we likely would have observed significant differences, with stride frequency increasing with the addition of load. Decreases in stride length would demand an increase in stride frequency to maintain the same marching rate.

The results for stance time, double support time and flight time are all consistent with previous experimental evidence.(Brown et al., 2014; E. Harman et al., 2000; Kinoshita, 1985) For the WK and RN trials, increases in load significantly increased the duration of stance phase. These effects were more pronounced during the RN trial, were a significant difference was observed between all load conditions. However, load had no significant effect on stance time during FM. Double support time during WK and FM trials significantly increased with increases in load and flight time during RN significantly decreased with increases in load. The levels of significance for the effect of load on spatiotemporal metrics are to be interpreted with some caution since the prescribed velocities at each load condition were slightly different. However, we can safely accept the overall trends within our findings, as they agree with previous studies that controlled for velocity. Barring the results for stride frequency, all other results for spatiotemporal metrics support our hypothesis for specific aim 1.

5.2 Kinematics

Since the effects of load carriage on lower limb kinematics were nearly bilaterally symmetrical, both left and right leg kinematics are discussed together. Any disparities will be highlighted in the subsections below.

5.2.1 Heel Strike

Results for knee joint angle support our hypothesis for specific aim 1 in that the knee was more flexed at heel strike with the addition of load. However, no significant increase in ankle dorsiflexion or hip flexion occurred in response to load as we hypothesized would happen. The knee was the only joint significantly affected by load at heel strike. The results showed that the knee joint was significantly more flexed during the loaded conditions when compared to BW. In a similar study, Kinoshita et al. investigated the effects of load carriage on select biomechanics of walking gait in male subjects. They also observed increased knee flexion with heavier loads. This was proposed to be a shock absorbing mechanism to reduce impact forces and allow for a smoother transfer of weight to the supporting surface.(Kinoshita, 1985) However, Tilbury-Davis and Hooper found no such phenomena with increasing loads.(Tilbury-Davis & Hooper, 1999) Their subjects were active duty military personnel who were regularly engaged in load carriage tasks. The difference in findings between the studies may be attributed to the dissimilarity in task familiarization of the participants. Those with load carriage experience likely have stronger lower extremity musculature and more developed patterns of movement as a result of training. The experienced warfighter will likely exhibit different, and possibly more efficient, biomechanical characteristics in response to load. These findings underscore the importance of load carriage training. As with any unfamiliar task, a periodized training plan employing the principles of specificity and progressive overload is essential to optimizing performance and reducing the risk of injury.

To no surprise, a significant main effect of locomotion was seen for all joint angles at heel strike. The Hip angle was significantly more flexed during FM compared to WK and RN. This increase in hip flexion contributed to the increase in stride length. Knee and ankle angles were significantly different for all locomotion patterns with the greatest knee more flexion and the ankle dorsiflexion observed during RN. Increased knee flexion at heel strike during running is a very common strategy for absorbing the impact forces and aiding in the storage of potential energy.(C. T. Farley & D. P. Ferris, 1998; Segers, 2006)

5.2.2 Mid Stance

Results for knee and ankle angle support our hypothesis for specific aim 1 in that the knee was more flexed and ankle more dorsiflexed at mid stance with the addition of load. However, no significant increase in hip flexion occurred in response to load as we hypothesized would happen. Previous studies that observed an increase in hip flexion with increasing load used a backpack load carriage system.(E. Harman et al., 2000; E. A. Harman et al., 2001; E. A. Harman et al., 2000; J. Knapik et al., 2010; Majumdar et al., 2010) The increase in hip flexion in those studies was attributed to the forward lean of the trunk to shift the CoM over the feet. (E. Harman et al., 2000; E. A. Harman et al., 2001; E. A. Harman et al., 2000; J. Knapik et al., 2010; Majumdar et al., 2010) The current study used a weight vest that distributed the load from front to back of the carrier evenly. Other studies using load carriage systems that evenly distributed the load on the carrier found no such increases in trunk lean or hip flexion.(Stewart A. Birrell & Haslam, 2010; Everett A. Harman, Frykman, Knapik, & Han, 1994; Kinoshita, 1985; J. Knapik et al., 2010; Seay, 2015) A significant main effect of load was seen again on knee angle at mid stance. The results for left and right knee were not the identical. The left knee was significantly more flexed under both loaded conditions compared to BW. The right knee, however, showed significantly more flexion only at +45%BW when compared to BW with no significant difference observed between +25%BW and BW. There was a significant interaction between the effects of load and locomotion on both left and right ankle angle, however, these effects were not bilaterally symmetrical. Load did not have the same effect on ankle angle at all locomotion conditions. For the left leg during both WK and FM, the ankle was more dorsiflexed with +45%BW when compared to BW and +25%BW. For the right leg during WK, the ankle was more dorsiflex with +45%BW when compared to +25%BW. Additionally, during RN, the angle was more dorsiflexed under both load conditions compared to BW. For both the left and right leg during FM, the ankle was more dorsiflexed with +45%BW compared to BW, however, no differences were observed between BW and +25%BW. The overall kinematic response to the addition of load at midstance was increased knee flexion and ankle dorsiflexion. As discussed in the previous subsection, a more flexed knee and dorsiflexed ankle is a means to absorb some of the impact forces and aid in the storage of potential energy.(C. T. Farley & D. P. Ferris, 1998; Segers, 2006) Additionally, increasing knee flexion lowers the body's CoM and improves stability during walking and running. The subtle contrast in the effect of load between left and right leg may be a notable finding worth further investigation. All subjects in the current study reported to be right leg dominant. The difference between dominant and non-dominant limb kinematics may be exacerbated with the onset of fatigue.

Once again, all joints were significantly affected by locomotion pattern. All locomotion patterns elicited significantly different hip angles at mid stance. The hip was most flexed during RN and most extend during WK. The knee was significantly more flexed, and the ankle dorsiflexed during RN when compared to WK and FM trials. These observations make sense when considering the characteristic dynamics of running. The dynamics of running are often compared to a bouncing ball. Running is characterized by an in-phase organization of potential energy and kinetic energy. (Kinoshita, 1985)'(Segers, 2006) The fluctuations of potential and kinetic energy share a more direct correlation and a considerable amount of energy is recovered through storage and release of energy from the elastic properties of the various muscles, tendons, and ligaments acting to carry out the movement. The more flexed position observed across all joints is an act of storing potential energy in preparation for its release during the transition from mid stance to toe off.

5.2.3 Toe Off

Results for knee joint angle support our hypothesis for specific aim 1 in that the knee was more flexed at toe off with the addition of load. However, no significant increase in ankle dorsiflexion or hip flexion occurred in response to load as we hypothesized would happen. There was a significant interaction between the effects of load and locomotion on knee angle at toe off. The interaction effect was slightly different for left and right leg. For the left leg, the simple effects of load were the same across locomotion patterns. The knee was most flexed with +45%BW and most extended at BW. The effect was nearly identical on the right leg, however, during FM no difference was observed between knee angle at toe off with +25%BW and +45%BW. Load affected knee and ankle angle at mid stance, but only affected knee angle and heel strike and toe off. The increased knee flexion with load is likely a factor contributing to reduced stride length with load. A more flexed knee at heel strike and toe off will result in less ground being covered with each stride.

All joints were significantly affected by locomotion pattern at toe off. The hip was significantly more flexed during RN and FM when compared to WK. The knee was significantly more flexed during RN when compared to WK with +25%BW. With +45%BW, the knee was significantly more flexed during RN when compared to both WK and FM. However, locomotion pattern had no significant effect on knee joint angle at BW. Ankle angle at BW and +45%BW was significantly different for all locomotion patterns. The ankle was most dorsiflexed during RN and most plantar flexed during WK. At the +25%BW condition, the ankle was significantly more dorsiflexed during RN, however, there was no difference between WK and FM. Again, these findings align with previous studies on running mechanics.(Brown et al., 2014; C. T. Farley & D. P. Ferris, 1998)

5.3 Kinetics

The majority of outcomes for kinetics resulted in an interaction between the effects of load and locomotion. Since the effects of load carriage on lower limb kinetics were nearly bilaterally symmetrical, both left and right leg kinetics are discussed together. Any disparities will be highlighted in the subsections below.

5.3.1 Heel Strike

Results for joint moment at heel strike do not fully support our hypothesis for specific aim 2. We hypothesized that increases in load would result in subsequent increases in relative joint moment at heel strike. There were no significant main effects of load averaged across all levels of locomotion. In other words, the effect of load on relative joint moment was not the same for each locomotion trial. Our outcomes for relative joint moment during a walking gait are not in complete agreement with previous research. Harman et al. found that absolute joint moment about the ankle, knee and hip significantly increased with load throughout the stride.(E. Harman et al., 2000) In the current study, a trend of increased absolute joint moments was observed with increases in load, however, the relative joint moments were seen to be significantly lower during certain locomotion trials. The relative hip extensor and knee flexor moments were significantly lower with +45%BW when compared to BW and +25%BW load conditions during WK and FM. The heaviest load for both the WK and FM locomotion trials elicited the lowest relative joint moments. Conversely, Wang et al. observed significant increases in the relative moments at the hip and knee and a trend of increased ankle dorsiflexion moment during a load carriage task with 32kg rucksacks.(Wang et al., 2013) However, the subjects recruited in their study were male participants. This may be an indication of the different biomechanical strategies employed by men and women in response to load carriage. Additionally, their protocol utilized a rucksack whereas the current study used a weight vest that evenly distributed the load from front to back of the carrier. Gait kinematics during backpack load carriage are different from the kinematics of load carriage with an evenly distributed load.(J. Knapik et al., 2010) This change in kinematics will in turn influence the moment experienced at each joint. Furthermore, in the Wang et al. study, marching velocity was the same between unloaded and loaded conditions.(Wang et al., 2013) As previously discussed, the velocities for each load condition in the current study were calculated percentages of the subject's GTP for that load condition. Since subjects transitioned to running at a slower velocity with load, the calculated treadmill velocity during WK and FM trials were slower with load. A decrease in marching velocity may have also contributed to the lower relative joint moments seen with +25%BW and +45%BW. These findings would suggest that joint moment can be effectively reduced by decreasing the rate of the march. Lastly, there was a significant increase in knee flexion moment with the addition of load during RN in the current study. This outcome supports our hypothesis and is in agreement with previous research. Brown et al. investigated the effects of load carriage on running biomechanics and observed similar increases in knee flexion moment.(Brown et al., 2014) The increased knee flexion moment at heel strike requires an equal and opposite moment to decelerate the body's CoM following heel strike. Brown et al. attributed this deceleration to the eccentric contraction of the knee extensors.(Brown et al., 2014) While it's reasonable to postulate that the knee extensors are contributing to deceleration, there may be several other tissues and structures within the joint acting to decelerate the body at HS. It cannot be said for sure that the knee extensors are the primary contributors. The effects of locomotion on joint moment were similar at each level of load. Relative joint moments at the hip and knee were significantly greater during FM at heel strike.

5.3.2 Mid Stance

Results for relative joint moment at mid stance do support our hypothesis for specific aim 2. We hypothesized that increases in load would result in subsequent increases in relative joint moment at heel strike. There were no significant main effects of load averaged across all levels of locomotion. In other words, the effect of load on relative joint moment was not the same for each locomotion trial. Hip flexion moment was significantly greater at BW than with either additional load during RN. Again, the lower velocities during the loaded conditions may be contributing the lower relative hip moments observed. Since higher knee joint moments were observed at heel strike during RN with the addition of load, the body may have experienced a faster deceleration prior to mid stance, possibly contributing to the lower moments observed at mid-stance. Knee flexion moment was significantly greater at BW during WK. Conversely, at BW during RN, there was a significantly greater knee extension moment. The difference in joint moment between WK and RN locomotion trials makes sense when considering the biomechanics of walking a running.(C. T. Farley & D. P. Ferris, 1998) During walking, the stance limb is more extended, as seen in our results. The knee flexors are likely contributing a significant amount of force to achieve a more extended limb position at mid stance while preparing for the second half of stance phase. During running, however, the knee is in a more flexed position throughout stance phase, also seen in our results. Starting at heel strike, the knee experiences a flexion moment as the body decelerates through the first half of stance phase.(C. T. Farley & D. P. Ferris, 1998) By mid stance, the transition to an extension moment would indicate an initiation of the release of stored elastic energy in preparation for toe off. As seen at the hip, joint moment at the knee was significantly lower with +25%BW and +45%BW, which does no support our hypothesis or previous research. Again, this may be further evidence of the difference in the biomechanical strategies utilized by men and women in response to load carriage. Additionally, as mentioned in previous sections, the lower marching velocities used for the loaded conditions may have also contributed to the lower relative joint moments observed in our results. A significant effect of load on ankle moment was only seen at midstance. Similar to the effects of load carriage on hip and knee moment, ankle plantarflexion moment was significantly greater at BW during RN for both left and right ankle. The right ankle also exhibited a significant main effect of load during WK with BW eliciting the greatest plantarflexion moment. Again, the effects of locomotion on joint moment were similar at each level of load. In general, all joint moments were significantly greater during RN at mid stance.

5.3.3 Toe Off

Once again, results for relative joint moment at toe off do support our hypothesis for specific aim 2. We hypothesized that increases in load would result in subsequent increases in relative joint moment at heel strike. At the left hip there was a significant main effect of load average across all levels of locomotion. Left hip extension moment was significantly greater at BW than with +45%BW. Overall, the right hip exhibited similar outcomes. However, during RN, right hip extension moment was greater at BW than with both +25%BW and +45%BW load conditions, and during FM, extension moment was greater at BW and +25%BW than with +45%BW. Both left and right knee flexion moment during WK was significantly greater at BW and +25%BW than with +45%BW. The right knee also presented effects of load during RN and FM. During RN, right knee flexion moment was significantly greater with both +25%BW and +45%BW than at BW. Results for right knee moment during FM were identical to WK, with a significantly greater flexion moment at BW and +25%BW than with +45%BW. Again, some of the bilateral asymmetries seen in the effect of load on lower limb kinematics and kinetics in our results, however subtle, are intriguing observations. Most studies investigating the effects of load carriage on gait biomechanics assess only one limb and assume the effects are bilateral symmetrical. Asymmetries across the body have been linked to an increased prevalence of injury in sport.(Brockett, Morgan, & Proske, 2004; Brughelli, Cronin, Mendiguchia, Kinsella, & Nosaka, 2010; Hewett, Lindenfeld, Riccobene, & Noyes, 1999) There may be some associated risk of injury for individuals with greater asymmetries during load carriage that has yet to be discovered. Overall, the effect of load on joint moments at toe off were similar to the result at heel strike and mid stance. Apart from a few differences between left and right limb, joint moments at the hip and knee were greater at BW. Again, the disagreement between our results and previous research is most likely a result of the disparity in the methodological approach. Previous studies looking at the effects of load carriage on gait kinetics controlled for marching velocity, whereas the current study used percentages of the GTP for each load condition. Once again, the effects of locomotion on joint moment were similar at each level of load. In general, joint moments at the hip and knee were greater during FM.

5.4 Coordination

Our observations for the effect of load magnitude and marching velocity on coordination do not fully support our hypotheses for specific aims 2 and 3. For specific aim 2, we hypothesized that increases in load carriage magnitude would decrease lower limb coordination variability. For specific aim 3, we hypothesized that coordination variability would be lower during RN compared to FM. Two exemplar subjects are presented for CRP analysis. Subject 04 experienced the overall lowest relative knee joint moments. Increases in load did not appear to decrease the variability of shank-thigh coupling for Subject 04. Additionally, there was no apparent difference in the variability of shank-thigh coupling between RN and FM conditions. However, there appeared to be less variability in shank-thigh coupling of the right leg compared to the left. Subject 06 experienced the overall highest relative knee joint moments. Increases in load appeared to decrease variability of shank-thigh coupling for the left leg, however, this trend did not appear to hold true for the right leg. There was no discernable difference in the variability of shank-thigh coupling between RN and FM conditions. Similar to subject 06, there appeared to be less variability in shank-thigh coupling of the right leg compared to the left. Overall Subject 04 appeared to have greater variability in shank-thigh coupling than subject 06. These observations may be an indication that, considering the lower relative joint moment experienced, subject 04 was better able to dissipate the load with greater variability in movement. Decreases in coordination variability have been identified as a potential factor for increases in mechanical stress.(Hamill, van Emmerik, Heiderscheit, & Li, 1999; Pollard, Heiderscheit, van Emmerik, & Hamill, 2005) The decrease of variability in shank-thigh coupling observed in subject 06 could be an indication that the same tissues and structures between the segments were repeatedly enduring the same mechanical stress from stride to stride. This repetitive stress, in combination with the higher relative knee joint moments, could increase the risk of lower extremity injuries. The fact that no differences in variability between load conditions within subject were observed could be attributed to the treadmill velocity. The lower treadmill velocities with the loaded conditions may not have been enough to elicit an effect on coordination variability. Much to our surprise, there was also no apparent difference in shank-thigh coordination variability between RN and FM locomotion trials. However, there appeared to be a bilateral asymmetry in coordination variability which seemed more evident with load during RN and FM conditions. Similarly, Diedrich and Warren found that coordination variability between left and right leg joints increased when subjects were forced to walk at velocities suitable for running.(Diedrich & Warren, 1995)

Yen et al. examined the effects of load carriage on trunk-thigh coordination variability and found that, contrary to their hypothesis, increases in load increased coordination variability.(Yen et al., 2012) However, they noticed a trended of decreasing variability with each additionally trial of data collection.(Yen et al., 2012) This suggest that the subject may not have been properly familiarized with the loads and marching velocities. Previous research on coordination of limb movement in novel tasks found that variability increased in subjects performing tasks that they were inexperienced in.(Wilson, Simpson, van Emmerik, & Hamill, 2008) The current study included an entirely separate day for load carriage familiarization in which subjects had 20 minutes of walking and jogging around their GTP under each load condition. Additionally, on the day of testing, immediately prior to data collection for each load condition, subjects performed 3 ramped treadmill protocols to establish their GTP. Seeing that our results show no apparent increase in variability with increases in load, subjects may have been properly familiarized prior to data collection.

5.5 Limitations and Future Research

This study has a few potential limitations. The desired sample size was not achieved. Data collection was postponed due to equipment failure. Only 15 subjects of the required 18 were enrolled at the time of equipment failure. Of those 15 that were enrolled, one subject withdrew, and another was unable to start testing before equipment failure, leaving 13 subjects that had undergone load carriage testing. Data from one of the 13 was of very low quality due to severe marker occlusion and the subject's inability to keep each foot on its respective left and right treadmill belt. This subject's data was therefore removed, and the data from a total of 12 subjects remained for analysis.

Furthermore, only female subjects were recruited for the present study. We cannot assume that male subjects would respond the same way to increases in load carriage magnitude during walking, running, and forced-marching. Many studies have identified differences in biomechanics and injuries sustained during load carriage between males and females. (Allison et al., 2015; B. Jones & Cowan, 1993; Martin & Nelson, 1986; R. M. Orr & Pope, 2016; Pandorf, Harman, Frykman, Obusek, & Smith, 1999) Additionally, with the exception of one subject, the current study used a civilian population with no formal military load carriage training and therefore the results cannot be extrapolated to a military population. Future research should include both males and females for comparison. Additionally, research investigating the effects of load carriage on the biomechanics of trained versus untrained males and females and exploring the prospective effects of load carriage training would be of interest.

Lastly, the current study used a cross-sectional design, comparing the biomechanical gait characteristics between load carriage conditions across marching velocities at similar points in time. The results are exclusive to each individuals current physical state. No conclusions can be drawn about how changes in physical state, such as fatigue, impact the effects of load carriage on gait biomechanics. However previous studies have shown that fatigue does significantly impact load carriage biomechanics. (Wang, Frame, Ozimek, Leib, & Dugan, 2012; Wang et al., 2013)

5.6 Conclusion

The purpose of this study was to investigate the effects of load carriage on female gait biomechanics. For specific aim 1, it was hypothesized that increases in load carriage magnitude would increase stride width, stride frequency, and double support time and decrease stride length and flight time at heel-strike, mid-stance, and toe-off across all levels of locomotion. With the exception of stride frequency, results for all other spatiotemporal metrics support our hypothesis. There were no interactions or significant main effects of load on stride frequency. However, the velocities for each load condition in the current study were calculated percentages of the subject's GTP for that load condition. Since treadmill velocity was lower for the loaded conditions, subjects did not need to adjust their stride frequency. Additionally, in specific aim 1, we hypothesized that increases in load carriage magnitude would increase the degree of joint flexion at heel-strike, mid-stance, and toe-off across all levels of locomotion. Our results only partially support our hypothesis. Results showed that knee flexion increased with increases in load at each gait event across all levels of locomotion. Ankle dorsiflexion increased with increases in load, but only at mid-stance. However, there were no interactions or significant main effects of load on hip angle at any gait event. This is likely a result of the weight vests evenly distributed load in our study. Other studies using load carriage systems that evenly distributed the load on the carrier found no such increases in trunk lean or hip flexion.(Stewart A. Birrell & Haslam, 2010; Everett A. Harman et al., 1994; Kinoshita, 1985; J. Knapik et al., 2010; Seay, 2015) Apart from hip flexion at heel-strike, all joints were more flexed during running at each gait event. Hip flexion was greater during forced-marching at heel-strike. For specific aim 2, it was hypothesized that increases in load carriage magnitude would increase relative joint moment at heel-strike, mid-stance and toe-off across all levels of locomotion. A preponderance of our results do not support our hypothesis.

Results showed that hip and knee moments decreased with increases in load during walking and forced-marching and increased during running at heel strike. Hip and knee moments decreased with increases in load at mid-stance and toe-off. Ankle moment decreased with increases in load at mid-stance only. There were no significant interactions or main effects of load on ankle moment at heel-strike or toe-off. These findings do not align with previous research. Wang et al. observed increases in relative joint moment with increases in load. However, treadmill velocity was constant across load conditions in their study. As previously discussed, the velocities for each load condition in the current study were calculated percentages of the subject's GTP for that load condition. Since subjects transitioned to running at a slower velocity with load, the calculated treadmill velocity during walk and forced-march trials were slower with load. A decrease in marching velocity may likely contributed to the lower relative joint moments seen with +25%BW and +45%BW. These findings would suggest that joint moment can be effectively reduced by decreasing the rate of the march. Hip and knee moments were greater during forced-marching at heel-strike and toe-off. However, all joint moments were greater during running at mid-stance. There were no significant main effects of locomotion on ankle moment at heel-strike or toe-off.

For specific aim 3, we hypothesized that increases in load carriage magnitude would decrease the variability in lower limb coordination. Our observations for the coordination variability of shank-thigh segments do support our hypothesis. Increases in load did not appear to decrease the variability of shank-thigh coupling. The lower treadmill velocities with the loaded conditions may not have been enough to elicit an effect on coordination variability. However, a difference between one subject who experienced overall greater joint moments and another who experienced lower joint moments was observed. The subject with greater joint moments appeared to have lower coordination variability compared the subject with lower joint moments. Decreases in coordination variability have been identified as a potential factor for increases in mechanical stress.(Hamill et al., 1999; Pollard et al., 2005) This repetitive stress, in combination with the higher relative knee joint moments, could increase the risk of lower extremity injuries. Additionally, there appeared to be a bilateral asymmetry in coordination variability which seemed more evident with load during RN and FM conditions. There may be some associated risk of injury for individuals with greater asymmetries during load carriage that has yet to be discovered. For specific aim 4, it was hypothesized that lower limb coordination patterns during running would have greater variability than during forced-marching. However, there was no apparent difference in shank-thigh coordination variability between RN and FM locomotion trials. This is one of the first studies to investigate the effects of combat load carriage on lower limb biomechanics and coordination variability during walking and running in female subjects. Some of the results from the current study support our hypotheses and findings from previous research, however, there were several outcomes that did not. These findings provide a basis for future research and express the importance of investigating gender differences in combat load carriage for military populations.

Bibliography

Abe, D., Muraki, S., & Yasukouchi, A. (2008). Ergonomic effects of load carriage on energy cost of gradient walking. *Applied Ergonomics, 39*(2), 144-149. doi:10.1016/j.apergo.2007.06.001

Abe, D., Yanagawa, K., & Niihata, S. (2004). Effects of load carriage, load position, and walking speed on energy cost of walking. *Applied Ergonomics, 35*(4), 329-335. doi:10.1016/j.apergo.2004.03.008

ACDC. (1962). A Study to Reduce the Load of the Infantry Combat Soldier. *US Army Combat Developments Command*.

ACDC. (1964). A Study to Conserve the Energy of the Combat Infantryman. *US Army Combat Developments Command*.

ADEA. (1987). Report on the ADEA Soldier's Load Initiative. *US Army Development and Employment Agency*.

Allison, K. F., Keenan, K. A., Sell, T. C., Abt, J. P., Nagai, T., Deluzio, J., . . . Lephart, S. M. (2015). Musculoskeletal, biomechanical, and physiological gender differences in the US military. *U.S. Army Medical Department journal*, 22.

Army, U. (1990). Foot Marches. *Field Manual*, 21-18.

Attwells, R. L., Birrell, S. A., Hooper, R. H., & Mansfield, N. J. (2006). Influence of carrying heavy loads on soldiers' posture, movements and gait. *Ergonomics, 49*(14), 1527-1537. doi:10.1080/00140130600757237

Bailey, T. L., & McDermott, W. M. (1952). *Review of research on load carrying*. Retrieved from

Birrell, S. A., & Haslam, R. A. (2009). The effect of military load carriage on 3-D lower limb kinematics and spatiotemporal parameters. *Ergonomics, 52*(10), 1298-1304. doi:10.1080/00140130903003115

Birrell, S. A., & Haslam, R. A. (2010). The effect of load distribution within military load carriage systems on the kinetics of human gait. *Applied Ergonomics, 41*(4), 585-590. doi:10.1016/j.apergo.2009.12.004

Bloom, D., & Woodhull-Mcneal, A. P. (1987). Postural adjustments while standing with two types of loaded backpack ∗. *Ergonomics, 30*(10), 1425-1430. doi:10.1080/00140138708966036

Bobet, J., & Norman, R. W. (1984). Effects of load placement on back muscle activity in load carriage. *European journal of applied physiology and occupational physiology, 53*(1), 71-75. doi:10.1007/BF00964693

Boling, M. C. (2009). A prospective investigation of biomechanical risk factors for patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort. *Am J Sports Med, 37*(11), 2108-2116. doi:10.1177/0363546509337934

Brennan, J. F. H., Jackson, C. R., Olsen, C., & Wilson, C. (2012). Blisters on the battlefield: the prevalence of and factors associated with foot friction blisters during Operation Iraqi Freedom I. *Mil Med, 177*(2), 157-162. doi:10.7205/MILMED-D-11-00325

Brockett, C. L., Morgan, D. L., & Proske, U. (2004). Predicting hamstring strain injury in elite athletes. *Med Sci Sports Exerc, 36*(3), 379-387. doi:10.1249/01.MSS.0000117165.75832.05

Brown, T. N., O'Donovan, M., Hasselquist, L., Corner, B. D., & Schiffman, J. M. (2014). Body borne loads impact walk-to-run and running biomechanics. *Gait & posture, 40*(1), 237-242. doi:10.1016/j.gaitpost.2014.04.001

Brughelli, M., Cronin, J., Mendiguchia, J., Kinsella, D., & Nosaka, K. (2010). Contralateral Leg Deficits in Kinetic and Kinematic Variables During Running in Australian Rules Football Players With Previous Hamstring Injuries. *J Strength Cond Res, 24*(9), 2539-2544. doi:10.1519/JSC.0b013e3181b603ef

Bush, R. A., Brodine, S. K., & Shaffer, R. A. (2000). The association of blisters with musculoskeletal injuries in male marine recruits. *Journal of the American Podiatric Medical Association, 90*(4), 194.

Caron, R. R., Lewis, C. L., Saltzman, E., Wagenaar, R. C., & Holt, K. G. (2015). Musculoskeletal stiffness changes linearly in response to increasing load during walking gait. *J Biomech, 48*(6), 1165-1171. doi:10.1016/j.jbiomech.2014.12.046

Cavanagh, P. R., & Williams, K. R. (1982). The effect of stride length variation on oxygen uptake during distance running. *Med Sci Sports Exerc, 14*(1), 30-35. doi:10.1249/00005768-198201000-00006

Coopers, D. (1981). *Research into foot lesions among Canadian Field Forces.* Paper presented at the 13th Commonwealth Defence Conference on Operational Clothing and Combat Equipment, Malaysia, CDA-11.

Dalén, Å., Nilsson, J., & Thorstensson, A. (1978). *Factors influencing a prolonged foot march*.

Dean, C. E. (2008). The Modern Warrior's Combat Load - Dismounted Operations in Afghanistan: 356. *Medicine & Science in Sports & Exercise, 40*(Supplement), 60. doi:10.1249/01.mss.0000321238.28174.55

Diedrich, F. J., & Warren, W. (1998). Dynamics of human gait transitions. *Timing of behavior: Neural, psychological, and computational perspectives*, 323-343.

Diedrich, F. J., & Warren, W. H. (1995). Why Change Gaits?: Dynamics of the Walk-Run Transition. *Journal of Experimental Psychology: Human Perception and Performance, 21*(1), 183-202. doi:10.1037/0096-1523.21.1.183

Diedrich, F. J., & Warren, W. H. (1998). The Dynamics of Gait Transitions: Effects of Grade and Load. *Journal of Motor Behavior, 30*(1), 60-78. doi:10.1080/00222899809601323

Drain, J., Orr, R., Attwells, R., & Billing, D. (2012). *Load Carriage Capacity of the Dismounted Combatant-A Commanders' Guide*. Retrieved from

Fallowfield, J. L., Blacker, S. D., Willems, M. E. T., Davey, T., & Layden, J. (2012). Neuromuscular and cardiovascular responses of Royal Marine recruits to load carriage in the field. *Applied Ergonomics, 43*(6), 1131. doi:10.1016/j.apergo.2012.04.003

Farley, C. T., & Ferris, D. P. (1998). Biomechanics of Walking and Running: Center of Mass Movements to Muscle Action. *Exercise and sport sciences reviews, 26*, 253-285.

Farley, C. T., & Ferris, D. P. (1998). Biomechanics of walking and running: center of mass movements to muscle action. *Exercise and sport sciences reviews, 26*, 253.

Feuerstein, M., Berkowitz, S. M., & Peck, J. C. A. (1997). Musculoskeletal-Related Disability in US Army Personnel: Prevalence, Gender, and Military Occupational Specialties. *Journal of Occupational & Environmental Medicine, 39*(1), 68-78. doi:10.1097/00043764-199701000-00013

Foch, E., & Milner, C. E. (2014). Frontal plane running biomechanics in female runners with previous iliotibial band syndrome. *J Appl Biomech, 30*(1), 58-65. doi:10.1123/jab.2013-0051

G*Power. (2009). G*Power (Version 3.1.9.2). Heinrish-Heine-Universitat Dusseldorf. Retrieved from http://www.gpower.hhu.de/en.html

Ghori, G. M., & Luckwill, R. G. (1985). Responses of the lower limb to load carrying in walking man. *European journal of applied physiology and occupational physiology, 54*(2), 145-150. doi:10.1007/BF02335921

Goh, J. H., Thambyah, A., & Bose, K. (1998). Effects of varying backpack loads on peak forces in the lumbosacral spine during walking. *Clin Biomech (Bristol, Avon), 13*(1 Suppl 1), S26-s31.

Grenier, J. G., Millet, G. Y., Peyrot, N., Samozino, P., Oullion, R., Messonnier, L., & Morin, J.-B. (2012). Effects of extreme-duration heavy load carriage on neuromuscular function and locomotion: a military-based study. *PloS one, 7*(8), e43586. doi:10.1371/journal.pone.0043586

Hamill, J., van Emmerik, R. E. A., Heiderscheit, B. C., & Li, L. (1999). A dynamical systems approach to lower extremity running injuries. *Clinical Biomechanics, 14*(5), 297-308. doi:10.1016/S0268-0033(98)90092-4

Hamilton, A. S., & Finklestein, H. E. (1944). MARCH FRACTURE: Report of a Case Involving Both Fibulae. *JBJS, 26*(1), 146-147.

Harman, E., Han, K.-H., Frykman, P., Johnson, M., Russell, F., & Rosenstein, M. (1992). THE EFFECTS ON GAIT TIMING, KINETICS, AND MUSCLE ACTIVITY OF VARIOUS LOADS CARRIED ON THE BACK: 774. *Medicine & Science in Sports & Exercise, 24*(Supplement), S129. doi:10.1249/00005768-199205001-00775

Harman, E., Hoon, K., Frykman, P., & Pandorf, C. (2000). *The effects of backpack weight on the biomechanics of load carriage*. Retrieved from

Harman, E. A., Frykman, P. N., Knapik, J. J., & Han, K.-H. (1994). 785 BACKPACK VS. FRONT-BACK PACK: DIFFERENTIAL EFFECTS OF LOAD ON WALKING POSTURE. *Medicine & Science in Sports & Exercise, 26*(Supplement), S140. doi:10.1249/00005768-199405001-00787

Harman, E. A., Han, K.-H., & Frykman, P. N. (2001). Load-Speed Interaction Effects on the Biomechanics of Backpack Load Carriage. *U.S. Army Research Institute of Environmental Medicine, ADP010991*, 76.

Harman, E. A., Han, K.-H., Frykman, P. N., & Pandorf, C. (2000). The Effects of Walking Speed on the Biomechanics of Backpack Load Carriage. *U.S. Army Research Institute of Environmental Medicine, Report No. T00-20*, 71.

Harper, W. H., Knapik, J. J., & De Pontbriand, R. (1997). *Female Load-Carrying Performance*. Retrieved from

Hauret, K. G., Jones, B. H., Bullock, S. H., Canham-Chervak, M., & Canada, S. (2010). Musculoskeletal injuries description of an under-recognized injury problem among military personnel. *Am J Prev Med, 38*(1 Suppl), S61-70. doi:10.1016/j.amepre.2009.10.021

Hearn, D., Rhon, D., Goss, D., & Thelen, M. (2017). Evaluation of a Novel Field Expedient Musculoskeletal Readiness Screening Tool in an Army Basic Training Population. *Mil Med, 182*(7), E1862. doi:10.7205/MILMED-D-16-00384

Hewett, T. E., Lindenfeld, T. N., Riccobene, J. V., & Noyes, F. R. (1999). The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study. *The American journal of sports medicine, 27*(6), 699.

Hreljac, A. (1995). Determinants of the gait transition speed during human locomotion: Kinematic factors. *J Biomech, 28*(6), 669-677. doi:10.1016/0021-9290(94)00120-S

Hughes, A., & Goldman, R. (1970). Energy cost of" hard work". *Journal of Applied Physiology, 29*(5), 570-572.

Johnson, R. C., Pelot, R. P., Doan, J. B., & Stevenson, J. M. (2000). *The Effect of Load Position on Biomechanical and Physiological Measures during a Short Duration March.* Paper presented at the Solider Mobility: Innovations in Load Carriage System Design and Evaluation, Kingston, Canada.

Jones, B., & Cowan, M. W. B. J. M. H. I. I. I. D. N. (1993). Intrinsic risk factors for exercise-related injuries among male and female army trainees. *The American journal of sports medicine, 21*(5), 705-710. doi:10.1177/036354659302100512

Jones, B., Hansen, B. C., Kaufman, K., & Brodine, S. (1996). Injuries in the military: a hidden epidemic. *Falls Church, VA: Armed Forces Epidemiological Board*.

Jones, B., & Knapik, J. (1999). Physical Training and Exercise-Related Injuries: Surveillance, Research and Injury Prevention in Military Populations. In (Vol. 27, pp. 111-125). NEW ZEALAND: Adis International.

Jones, B. H. (1983). Overuse injuries of the lower extremities associated with marching, jogging, and running: a review. *Mil Med, 148*(10), 783-787.

Jones, B. H., Amoroso, P., Canham, M., Weyandt, M., & Schmitt, J. (1999). Atlas of injuries in the US Armed Forces. *Mil Med, 164*(8S), 9-1.

Jones, B. H., Harris, J. M., Vinh, T. N., & Rubin, C. (1989). Exercise-induced stress fractures and stress reactions of bone: epidemiology, etiology, and classification. *Exercise and sport sciences reviews, 17*, 379.

Kaufman, K. R., Brodine, S., & Shaffer, R. (2000). Military training-related injuries: surveillance, research, and prevention. *Am J Prev Med, 18*(3 Suppl), 54-63. doi:10.1016/S0749-3797(00)00114-8

Keren, G., Epstein, Y., Magazanik, A., & Sohar, E. (1981). The energy cost of walking and running with and without a backpack load. *European journal of applied physiology and occupational physiology, 46*(3), 317-324. doi:10.1007/BF00423407

Kinoshita, H. (1985). Effects of different loads and carrying systems on selected biomechanical parameters describing walking gait. *Ergonomics, 28*(9), 1347-1362. doi:10.1080/00140138508963251

Knapik, J., Harman, E., & Reynolds, K. (1996). Load carriage using packs: A review of physiological, biomechanical and medical aspects. *Applied Ergonomics, 27*(3), 207-216. doi:10.1016/0003-6870(96)00013-0

Knapik, J., Johnson, R., Ang, P., Meiselman, H., & Bensel, C. (1993). *Road march performance of special operations soldiers carrying various loads and load distributions*. Retrieved from

Knapik, J., Montain, S., McGraw, S., Grier, T., Ely, M., & Jones, B. (2012). Stress Fracture Risk Factors in Basic Combat Training. *Int J Sports Med, 33*(11), 940-946. doi:10.1055/s-0032-1311583

Knapik, J., Reynolds, K., Santee, W. R., Friedl, K., & Borden, I. (2010). *Load carriage in military operations: a review of historical, physiological, biomechanical, and medical aspects*. Washington, D.C. U6 US U7 - eBook: Borden Institute.

Knapik, J., Reynolds, K., Staab, J., Vogel, J. A., & Jones, B. (1992). Injuries associated with strenuous road marching. *Mil Med, 157*(2), 64-67.

Knapik, J., Staab, J., Bahrke, M., Reynolds, K., Vogel, J., & O'Connor, J. (1991). Soldier performance and mood states following a strenuous road march. *Mil Med, 156*(4), 197-200.

Knapik, J. J., Ang, P., Meiselman, H., Johnson, W., Kirk, J., Bensel, C., & Hanlon, W. (1997). Soldier performance and strenuous road marching: influence of load mass and load distribution. *Mil Med, 162*(1), 62-67.

Knapik, J. J., & Reynolds, K. (2016). Load Carriage-Related Injury Mechanisms, Risk Factors, and Prevention. In A. Gefen & Y. Epstein (Eds.), *The Mechanobiology and Mechanophysiology of Military-Related Injuries* (pp. 107-137). Cham: Springer International Publishing.

Knapik, J. J., Reynolds, K. L., & Harman, E. (2004). Soldier Load Carriage: Historical, Physiological, Biomechanical, and Medical Aspects. *Mil Med, 169*(1), 45-56.

Knapik, J. J., Sharp, M. A., Canham-Chervak, M., Hauret, K., Patton, J. F., & Jones, B. H. (2001). Risk factors for training-related injuries among men and women in basic combat training. *Med Sci Sports Exerc, 33*(6), 946-954. doi:10.1097/00005768-200106000-00014

Krupenevich, R., Rider, P., Domire, Z., & DeVita, P. (2015). Males and Females Respond Similarly to Walking With a Standardized, Heavy Load. *Mil Med, 180*(9), 994.

Kujala, U. M., Kvist, M., Osterman, K., Friberg, O., & Aalto, T. (1986). Factors predisposing Army conscripts to knee exertion injuries incurred in a physical training program. *Clinical Orthopaedics and Related Research*(210), 203-212. doi:10.1097/00003086-198609000-00029

LaFiandra, M., & Harman, E. (2004). The distribution of forces between the upper and lower back during load carriage. *Med Sci Sports Exerc, 36*(3), 460-467. doi:10.1249/01.MSS.0000117113.77904.46

LaFiandra, M., Wagenaar, R. C., Holt, K. G., & Obusek, J. P. (2003). How do load carriage and walking speed influence trunk coordination and stride parameters? *J Biomech, 36*(1), 87-95. doi:10.1016/S0021-9290(02)00243-9

Liew, B. X. W., Morris, S., & Netto, K. (2016). Joint power and kinematics coordination in load carriage running: Implications for performance and injury. *Gait & posture, 47*, 74-79. doi:10.1016/j.gaitpost.2016.04.014

Ling, W., Houston, V., Tsai, Y.-S., Chui, K., & Kirk, J. (2004). Women's Load Carriage Performance Using Modular Lightweight Load-Carrying Equipment. *Mil Med, 169*(11), 914-919.

Lohman, T. G., Harris, M., Teixeira, P. J., & Weiss, L. (2000). Assessing Body Composition and Changes in Body Composition: Another Look at Dual-Energy X-ray Absorptiometry. *Annals of the New York Academy of Sciences, 904*(1), 45-54. doi:10.1111/j.1749-6632.2000.tb06420.x

Lothian, N. (1921). The load carried by the soldier. *Journal of the Royal Army Medical Corps, 37*(4), 241-263.

Majumdar, D., Pal, M. S., & Majumdar, D. (2010). Effects of military load carriage on kinematics of gait. *Ergonomics, 53*(6), 782-791. doi:10.1080/00140131003672015

Mann, R. A., & Hagy, J. (1980). Biomechanics of walking, running, and sprinting. *The American journal of sports medicine, 8*(5), 345-350. doi:10.1177/036354658000800510

Martin, P. E., & Nelson, R. C. (1986). The effect of carried loads on the walking patterns of men and women. *Ergonomics, 29*(10), 1191-1202. doi:10.1080/00140138608967234

Mattila, V. M., Niva, M., Kiuru, M., & Pihlajamäki, H. (2007). Risk factors for bone stress injuries: a follow-up study of 102,515 person-years. *Med Sci Sports Exerc, 39*(7), 1061-1066. doi:10.1249/01.mss.0b013e318053721d

Mercier, J., Le Gallais, D., Durand, M., Goudal, C., Micallef, J. P., & Préfaut, C. (1994). Energy expenditure and cardiorespiratory responses at the transition between walking and running. *European journal of applied physiology and occupational physiology, 69*(6), 525-529. doi:10.1007/BF00239870

Murray, M. P. (1967). Gait as a total pattern of movement. *American journal of physical medicine, 46*(1), 290.

Nindl, B. C., Williams, T. J., Deuster, P. A., Butler, N. L., & Jones, B. H. (2013). Strategies for optimizing military physical readiness and preventing musculoskeletal injuries in the 21st century. *U.S. Army Medical Department journal*, 5.

Obusek, J. P., Harman, E. A., Frykman, P. N., Palmer, C. J., & Bills, R. K. (1997). THE RELATIONSHIP OF BACKPACK CENTER OF MASS LOCATION TO THE METABOLIC COST OF LOAD CARRIAGE 1170. *Medicine &amp Science in Sports &amp Exercise, 29*(Supplement), 205. doi:10.1097/00005768-199705001-01168

Orr, R. M., Johnston, V., Coyle, J., & Pope, R. (2015). Reported load carriage injuries of the Australian army soldier. *Journal of occupational rehabilitation, 25*(2), 316-322. doi:10.1007/s10926-014-9540-7

Orr, R. M., & Pope, R. (2016). Gender differences in load carriage injuries of Australian army soldiers. *BMC Musculoskelet Disord, 17*(1), 488. doi:10.1186/s12891-016-1340-0

Orr, R. M., Pope, R., Johnston, V., & Coyle, J. (2014). Soldier occupational load carriage: a narrative review of associated injuries. *International Journal of Injury Control and Safety Promotion, 21*(4), 388-396. doi:10.1080/17457300.2013.833944

Pandorf, C. E., Harman, E. A., Frykman, P. N., Obusek, J. P., & Smith, T. J. (1999). A COMPARISON OF THE LOAD CARRIAGE BIOMECHANICS OF MALE AND FEMALE SOLDIERS. *Medicine & Science in Sports & Exercise, 31*(Supplement), S190. doi:10.1097/00005768-199905001-00851

Patton, J. F., Kaszuba, J., Mello, R. P., & Reynolds, K. L. (1991). Physiological responses to prolonged treadmill walking with external loads. *European journal of applied physiology and occupational physiology, 63*(2), 89-93. doi:10.1007/BF00235175

Piantanida, N. A., Knapik, J. J., Brannen, S., & O'Connor, F. (2000). Injuries during Marine Corps officer basic training. *Mil Med, 165*(7), 515.

Pollard, C. D., Heiderscheit, B. C., van Emmerik, R. E. A., & Hamill, J. (2005). Gender differences in lower extremity coupling variability during an unanticipated cutting maneuver. *Journal of Applied Biomechanics, 21*(2), 143-152. doi:10.1123/jab.21.2.143

Popovich, R. M., Gardner, J. W., Potter, R., Knapik, J. J., & Jones, B. H. (2000). Effect of rest from running on overuse injuries in Army basic training. *Am J Prev Med, 18*(3), 147-155. doi:10.1016/S0749-3797(99)00167-1

Quesada, P. M., Mengelkoch, L. J., Hale, R. C., & Simon, S. R. (2000). Biomechanical and metabolic effects of varying backpack loading on simulated marching. *Ergonomics, 43*(3), 293-309. doi:10.1080/001401300184413

Renbourn, E. T. (1954). The knapsack and the pack: Part 3. *Journal of the Royal Army Medical Corps, 100*(2), 193-200.

Reynolds, K. L., Kaszuba, J., Mello, R. P., & Patton, J. F. (1990). *Prolonged treadmill load carriage: acute injuries and changes in foot anthropometry*. Retrieved from

Ruscio, B., Smith, J., Amoroso, P., Anslinger, J., Bullock, S., Burnham, B., . . . Garver, R. (2006). *DoD Military Injury Prevention Priorities Working Group: Leading injuries, causes and mitigation recommendations*. Retrieved from

Seay, J. F. (2015). Biomechanics of Load Carriage—Historical Perspectives and Recent Insights. *J Strength Cond Res, 29 Suppl 11*(11), S129-S133. doi:10.1519/JSC.0000000000001031

Segers, V. (2006). *A biomechanical analysis of the realization of actual human gait transition.* Ghent University,

Sell, T. C., Chu, Y., Abt, J. P., Nagai, T., Deluzio, J., McGrail, M. A., . . . Lephart, S. M. (2010). Minimal Additional Weight of Combat Equipment Alters Air Assault Soldiers Landing Biomechanics. *Mil Med, 175*(1), 41-47. doi:10.7205/MILMED-D-09-00066

Shaffer, R. A., Brodine, S. K., Almeida, S. A., Williams, K. M., & Ronaghy, S. (1999). Use of simple measures of physical activity to predict stress fractures in young men undergoing a rigorous physical training program. *American journal of epidemiology, 149*(3), 236-242. doi:10.1093/oxfordjournals.aje.a009797

Silder, A., Delp, S. L., & Besier, T. (2013). Men and women adopt similar walking mechanics and muscle activation patterns during load carriage. *J Biomech, 46*(14), 2522. doi:10.1016/j.jbiomech.2013.06.020

Soule, R. G., Pandolf, K. B., & Goldman, R. F. (1978). Energy expenditure of heavy load carriage. *Ergonomics, 21*(5), 373-381. doi:10.1080/00140137808931734

Strowbridge, N. F. (2002). Musculoskeletal injuries in female soldiers: analysis of cause and type of injury. *Journal of the Royal Army Medical Corps, 148*(3), 256-258. doi:10.1136/jramc-148-03-06

Stuempfle, K. J., Drury, D. G., & Wilson, A. L. (2004). Effect of load position on physiological and perceptual responses during load carriage with an internal frame backpack. *Ergonomics, 47*(7), 784-789. doi:10.1080/0014013042000193264

Sutton, E. L. (1976). PREPARING FOR COMBAT: ATHLETIC INJURIES INCURRED AND PERFORMANCE LIMITING ORTHOPEDIC AND MEDICAL CONDITIONS.

Medicine & Science in Sports & Exercise, 8(1), 74. doi:10.1249/00005768-197621000-00134

Swanik, C. B. (2015). Brains and Sprains: The Brain's Role in Noncontact Anterior Cruciate Ligament Injuries. *Journal of athletic training, 50*(10), 1100-1100. doi:10.4085/1062-6050-50.10.08

Taga, G. (1995a). A model of the neuro-musculo-skeletal system for human locomotion. I. Emergence of basic gait. *Biol Cybern, 73*(2), 97-111. doi:10.1007/BF00204048

Taga, G. (1995b). A model of the neuro-musculo-skeletal system for human locomotion. II Real-time adaptability under various constraints. *Biol Cybern, 73*(2), 113-121. doi:10.1007/BF00204049

Thelen, E., Ulrich, B. D., & Society for Research in Child, D. (1991). *Hidden skills: a dynamic systems analysis of treadmill stepping during the first year* (Vol. 56, no. 1.). Chicago: The University of Chicago Press.

Tilbury-Davis, D. C., & Hooper, R. H. (1999). The kinetic and kinematic effects of increasing load carriage upon the lower limb. *Human Movement Science, 18*(5), 693-700. doi:10.1016/S0167-9457(99)00026-3

Treloar, A. K. L., & Billing, D. C. (2011). Effect of load carriage on performance of an explosive, anaerobic military task. *Mil Med, 176*(9), 1027-1031. doi:10.7205/MILMED-D-11-00017

Turvey, M. T. (1990). Coordination. *American Psychologist, 45*(8), 938-953. doi:10.1037/0003-066X.45.8.938

Turvey, M. T., Holt, K. G., Obusek, J., Salo, A., & Kugler, P. N. (1996). Adiabatic transformability hypothesis of human locomotion. *Biol Cybern, 74*(2), 107-115.

Wang, H., Frame, J., Ozimek, E., Leib, D., & Dugan, E. L. (2012). Influence of fatigue and load carriage on mechanical loading during walking. *Mil Med, 177*(2), 152-156. doi:10.7205/MILMED-D-11-00210

Wang, H., Frame, J., Ozimek, E., Leib, D., & Dugan, E. L. (2013). The effects of load carriage and muscle fatigue on lower-extremity joint mechanics. *Res Q Exerc Sport, 84*(3), 305-312. doi:10.1080/02701367.2013.814097

Willems, P. A., Heglund, N. C., Cavagna, G. A., & Penta, M. (1995). Energy-saving gait mechanics with head-supported loads. *Nature, 375*(6526), 52-54. doi:10.1038/375052a0

Wilson, C., Simpson, S. E., van Emmerik, R. E., & Hamill, J. (2008). Coordination variability and skill development in expert triple jumpers. *Sports Biomech, 7*(1), 2-9. doi:10.1080/14763140701682983

Witvrouw, E., Lysens, R., Bellemans, J., Cambier, D., & Vanderstraeten, G. (2000). Intrinsic risk factors for the development of anterior knee pain in an athletic population. A two-year prospective study. *The American journal of sports medicine, 28*(4), 480.

Yen, S.-C., Gutierrez, G. M., Ling, W., Magill, R., & McDonough, A. (2012). Coordination variability during load carriage walking: Can it contribute to low back pain? *Human Movement Science, 31*(5), 1286-1301. doi:10.1016/j.humov.2012.02.006