SAGITTAL AND FRONTAL PLANE KNEE ANGULAR JERK EFFECTS DURING PROLONGED LOAD CARRIAGE

by

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DEDICATION

I would like to dedicate my thesis to my sister Valerie. You had much more of an impact on my life than you will ever realize, and it breaks my heart that you were unable to be there through the end. I am very thankful to have a sister who was so supportive and always wanted the best for me. Aside from our parents, I do not know if I had someone who bragged about me as much as you did. Thank you for always pushing me to achieve more.

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ABSTRACT

Introduction: Musculoskeletal injuries are a costly military problem that routinely occur during training. Quantifying smoothness of knee motion, or angular knee jerk, may be an effective measure to monitor injury risk during training, but to date, the effects of body borne load and prolonged locomotion on angular knee jerk are unknown. **Purpose:** This study sought to quantify angular knee jerk for frontal and sagittal plane motion during prolonged load carriage. **Methods:** Eighteen participants had peak and cost of angular jerk for frontal and sagittal plane knee motion quantified while they walked (1.3 m/s) 60-minutes with three body borne loads (0, 15, and 30 kg). **Statistical Analysis:** Peak and cost of angular jerk for sagittal and frontal plane knee motion of stance phase (0 % - 100%) were derived from motion capture and IMU data and submitted to a repeated measures linear model to test the main effects and interaction of load (*0, 15, and 30 kg*) and time (*0, 15, 30, 45, and 60 min.*). Two one sided t-tests (TOSTs) were used to compare the motion capture- and IMU-derived measures of angular jerk for sagittal and frontal plane knee motion. **Results:** For the motion capture-derived jerk measures, body borne load increased peak and cost of angular jerk for sagittal ($p < 0.001$, $p < 0.001$) and frontal ($p < 0.001$, $p < 0.001$) plane knee motion, while time increased jerk cost ($p = 0.001$) of frontal plane knee motion. While the IMU-derived jerk measures exhibited similar increases in peak and cost of angular jerk for sagittal ($p < 0.001$, $p < 0.001$) and frontal ($p = 0.027$, $p < 0.001$) plane knee motion with addition of load, and in cost ($p = 0.015$) of angular jerk for frontal plane knee motion with time, they were not statistically equivalent to motion-capture derived measures ($p > 0.05$).

Conclusion: Prolonged load carriage may lead to jerkier knee motion and increased knee musculoskeletal injury risk. Specifically, the jerkier knee motions exhibited with the addition of body borne load and longer walking time may increase the joint loading that leads to greater knee musculoskeletal injury risk.

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IMU	Inertial Measurement Unit

CHAPTER ONE: INTRODUCTION

Musculoskeletal injuries are a common and costly problem for the military. Up to 12% of military personnel suffer a musculoskeletal injury each month, resulting in over 2.4 million health care visits annually[1,2]. The Marine Corps alone spends about $111 million per year treating musculoskeletal injuries, yet still has 356,000 lost duty days annually due to these injuries[2,3]. A majority of these musculoskeletal injuries are overuse and occur at or below the knee during training activities[1,4,5]. During training activities, soldiers routinely carry body borne loads between 20 kg and 40 kg, which increase injury risk by altering lower limb neuromechanics[6–10]. Considering military load carriage is reportedly a risk for lower limb musculoskeletal injury in general, and knee musculoskeletal injury specifically, understanding knee neuromechanics during load carriage is imperative to successfully reduce the incidence and cost associated with these debilitating injuries[1,5,11].

During military training, load carriage is a risk factor for knee musculoskeletal injury[1,12,13]. During specific training activities, such as prolonged walking, the addition of body borne load produces alterations of knee biomechanics thought to stabilize the joint, but may increase hazardous loading of joint's soft tissues[14]. Specifically, when walking with body borne load, vertical ground reaction forces significantly increase, resulting in greater peak and range of knee flexion motion to help stabilize the joint[7,13–20]. Greater knee flexion reportedly leads to larger joint contact forces and loading on the knee's soft-tissue structures, increasing overuse injury risk[13–15,18,20]. Walking with body borne load also purportedly increases both the angle and magnitude of the mediolateral ground reaction force[15,19,21]. A larger and more medially-directed ground reaction force acts to push the knee into varus and increases the external knee adduction moment, which loads the medial knee joint compartment and decreases mediolateral joint stability[19,22,23]. Both greater peak knee adduction joint angle and moment predict medial joint compartment loading and are related to the progression of knee injury and pain, as well as joint musculoskeletal disease (i.e., osteoarthritis)[23–27]. Yet, despite the direct link to musculoskeletal injury, there is currently a dearth of information about frontal plane knee biomechanics (i.e., joint adduction angle and moment) during locomotion with body borne load.

Prolonged load carriage may lead to fatigue and increased knee musculoskeletal injury risk[28–30]. Fatigue, or failure to produce required muscular force to maintain joint stability, results in significant changes to knee biomechanics during locomotion. During prolonged locomotion without body borne load, individuals exhibit an increase in vertical ground reaction force, resulting in an increase in peak knee flexion angle and impulse[20,30–32]. These biomechanical changes are thought to increase knee joint loading and overuse injury risk[30]. When fatigue is combined with the addition of body borne load, alterations in knee biomechanics, such as peak knee flexion angle and moment, are reported to further increase[20,31–33]. However, to date there is limited information about the effects of fatigue in the frontal plane of knee motion.

Quantifying smoothness of knee motion may be an effective measure of joint instability and injury risk[34,35]. Angular jerk, the rate of change of acceleration, reportedly estimates the smoothness of a kinematic parameter. In fact, angular jerk cost may be the best way to quantify the smoothness of joint movement, as the movement trajectory with the smoothest motion also exhibits the lowest jerk cost[36–40]. Quantitatively, jerk cost is:

$$Jerk\ Cost = \frac{1}{2}\int_0^t \left(\frac{d^3\theta}{dt^3}\right)^2 dt,$$

where t is time (sec) and θ is knee angle (rad)[36]. A movement trajectory that minimizes jerk uses less energy to execute and places smaller loads on the joint, reducing risk of injury[34,35]. In patients with radiographically confirmed musculoskeletal disease, angular jerk cost of the knee increases in both the sagittal and frontal planes of motion[41,42]. However, currently the effects of load carriage and fatigue on angular knee jerk are unknown[36,43,44].

Traditionally, kinematic data are recorded with a motion capture system. However, motion capture systems are expensive, often limited to a laboratory setting, and have difficulty collecting data that represents day-to-day activities. Researchers have recently started using inertial measurement units (IMUs) to calculate kinematic data, as they are cheaper and can continuously collect acceleration-based data[45–48]. The main achievements for IMU-derived kinematic data have been to create algorithms to quantify sagittal plane joint kinematics, particularly at the knee[48–51]. Currently, however, IMU derived kinematics data are limited to sagittal plane, and it is relatively unknown how these measures compare to motion capture derived metrics. With that in mind, this study aims to fill that critical void, and quantify angular knee jerk for frontal and sagittal plane motion during prolonged load carriage with both IMU and motion capture systems.

Specific Aims

Specific Aim 1

To determine whether jerk of knee motion increases with the addition of body borne load. Specifically, this study will quantify peak and cost of angular jerk for frontal and sagittal plane knee motion during an over-ground walking task (1.3 m/s) with three body borne loads (0, 15, and 30 kg).

Hypothesis 1.1

Participants will exhibit a significant increase in peak and cost of angular jerk for sagittal plane knee motion with each incremental addition (0, 15, and 30 kg) of body borne load.

Hypothesis 1.2

Participants will exhibit a significant increase in peak and cost of angular jerk for frontal plane knee motion with each incremental addition (0, 15, and 30 kg) of body borne load.

Significance

Examining the jerk of sagittal and frontal plane knee motion may provide a quantitative way to observe the detrimental effects of body borne load during locomotor tasks. Understanding the effect of body borne load on sagittal and frontal plane knee jerk will provide the military the information necessary to decrease incidence of training-related knee musculoskeletal injuries and a quantitative way to measure effectiveness of injury prevention protocols.

Specific Aim 2

To determine whether the jerk of knee motion increases throughout the duration of a prolonged load carriage task. Specifically, this study will quantify peak and cost of angular jerk for sagittal and frontal knee motion starting at minute 0 and every 5 minutes thereafter, while participants walk (1.3 m/s) over-ground for 60 minutes with three different body borne loads (0, 15, and 30 kg).

Hypothesis 2.1

Participants will exhibit a significant increase in peak and cost of angular jerk for sagittal plane knee motion throughout the duration (15, 30, 45, and 60 min.) of the prolonged load carriage task.

Hypothesis 2.2

Participants will exhibit a significant increase in peak and cost of angular jerk for frontal plane knee motion throughout the duration (15, 30, 45, and 60 min.) of the prolonged load carriage task.

Significance

Determining if the sagittal and frontal plane knee motions are jerkier throughout the prolonged carriage task can help the military reduce the incidence of overuse knee injuries. The military can use this information to monitor injury risk during training and occupational related locomotor activities.

Specific Aim 3

To determine whether motion capture and accelerometer-derived measurements of angular jerk for sagittal and frontal plane knee motion are equivalent. Specifically, this study will quantify peak and cost of angular jerk for sagittal and frontal plane knee motion derived from both motion capture and accelerometer (IMU) data during a 60-minute over-ground walking task with three body borne loads and determine whether these measures are statistically equivalent.

Hypothesis 3.1

The IMU- and motion capture-derived peak and cost of angular jerk for sagittal plane knee motion will be statistically equivalent.

Hypothesis 3.2

The IMU- and motion capture-derived peak and cost of angular jerk for frontal plane knee motion will be statistically equivalent.

Significance

Determining whether accelerometer-derived measures on knee angular jerk are equivalent to the gold standard motion capture derived measurements will provide the military the ability to quantify knee biomechanics during ecologically valid settings. Specifically, the military will be able to collect kinematic data outside of the laboratory, during actual training, or operational activities.

CHAPTER TWO: LITERATURE REVIEW

The following section aims to detail knee biomechanics, specifically 1) musculoskeletal injury in the military, 2) load carriage 3) fatigue effects on biomechanics, 4) jerk in lower limb, and 5) inertial measurement units (IMUs) reliability in measuring knee kinematics.

Musculoskeletal Injury

Musculoskeletal injury occurs when the musculoskeletal system is damaged from physical trauma due to a large amount of energy being transferred to the tissue[52,53]. Factors such as age, sex, body composition, and activity level may lead to an individual being more likely to sustain one of these injuries, which are usually sustained in the lower limb[52,54]. These lower limb musculoskeletal injuries are thought to be brought on by an increase in varus/valgus and internal/external moments at the knee, which increase the loading at the knee by pushing the knee out of its anatomical alignment[55]. This increase in loading can lead to injuries such as anterior cruciate ligament (ACL) injury, which may result in an early onset of other musculoskeletal injuries, such as knee osteoarthritis[52].

In the Military

Musculoskeletal injuries are a common and costly problem for the military, being cited as the most important health problem the military is currently facing[9]. An estimated 900,000 service members are affected by musculoskeletal injuries annually, resulting in 2.4 million medical visits and an associated cost of $548 million[2]. The Marine Corps alone spends around $111 million per year on treating musculoskeletal injuries, yet has 356,000 limited duty days as a result of these injuries, which commonly happen during training[1,3,6]. Up to 12% of military recruits receive a musculoskeletal injury each month, with 78% of these injuries being overuse injuries, usually resulting from repetitive strenuous activity, namely running and conditioning hikes[1,3,6,11]. The high rate of these injuries can be attributed to repetitive strenuous activity, namely running and conditioning hikes, which result in 75% of all musculoskeletal injuries developing from cumulative microtrauma, likely from repetitive impact forces, and 82% of those injuries being in the lower limb[5,6]. Two major contributors to lower limb musculoskeletal injury during training are load carriage and fatigue[6,11].

Load Carriage

As technology has advanced, soldiers have started to carry heavier body borne loads, regularly carrying body borne loads between 20 and 40 kg[4,6,10,56]. These body borne loads are often overloaded, exceeding the recommended weight from the Army Field Manual of 32.7 kg, and are not ergonomically designed, leading to fatigue and poor load carriage[6]. The additional weight can reach up to 90% of a soldier's body weight, resulting in the alteration of lower limb neuromechanics, increasing the risk of musculoskeletal injury[3,6,7,15].

Effects on Knee and Ground Reaction Forces

Walking with body borne load affects the kinematics and kinetics of gait. Both vertical and sagittal ground reaction force increase with the presence of body borne load, with peak ground reaction force increasing significantly at as low as a 20 kg load, resulting in changes up the kinetic chain[13–15,17–19]. Previous studies have shown mixed results regarding the range of motion of sagittal plane knee range of motion while walking with body borne load, but knee flexion has consistently been shown to increase with the addition of body borne load to help stabilize the knee[7,12–14,18,19,22,56–58]. In order to stabilize the knee, knee flexion helps the body lower the center of mass, but this also increases the forces and torques on the knee, increasing loading on the soft tissues at the knee[7,12,14–16,18]. Additionally, this alteration of knee flexion produces the push off force of the foot, which is normally produced by the extension of the hip, increasing the strain of the muscles at the knee[59].

With loads greater than 30% of body weight, an increase in mediolateral impulse occurs, though there is conflicting evidence as to the effects of load on mediolateral kinematics[22,32]. Some studies show there are no kinematic mediolateral changes, but others show significant changes with load, namely that knee ab/adduction increases with the addition of body borne load, indicating the effect of gait compensations[12,15,19,22,60]. However, an increase in medial ground reaction force is consistently seen, indicating a shift in center of mass and instability[14,19,23]. With the increase in ground reaction force, knee adduction moment increases, shifting the knee into varus and increasing loading on the medial joint compartment, especially at the anterior cruciate ligament[19,23,26,27,61]. Increases in knee adduction moment and lower limb alignment, which are seen with the addition of body borne load, are significantly correlated with the onset of knee osteoarthritis[24,25,62]. Additionally, greater valgus position has been linked with an increase in patellofemoral pain[63]. Despite the links to musculoskeletal injury, there is a dearth of information on load carriage effects on frontal plane knee biomechanics.

Effect on Lower Limb

Along with knee kinematics, other lower limb changes occur with the addition of body borne load. An increase in load leads to an increase in ankle dorsiflexion during stance phase, along with an increase in hip flexion/extension range of motion[13,31]. Ankle and hip torque and joint forces both increase with an increase in load[14,18,22]. Along with kinematic and kinetic changes, spatiotemporal changes can be seen with the addition of body borne load. An increase in load has shown to lead to a decrease in stride length, thus increasing the time in double support during stance phase in order to help stabilize the body[7,13,18,57,59]. In another attempt to stabilize the body with the addition of body borne load, stride width also decreases and stride frequency increases[14,22]. With an increase in load weight, a correlating increase in physiological cost has also been seen[64]. An increase in VO_2 and heart rate can be seen with the addition of body borne load, indicating an increase in energy expenditure[13,18].

Because load carriage alters gait, the musculoskeletal system is placed under more stress[18,65]. With this alteration of gait, there is an associated increase in fall risk[31]. In an attempt to mitigate the risk of injury, body borne load is recommended to be carried as close to the center of mass as possible, preferably by a double pack or a backpack.[64,65]. With an increase in load, studies have found that the lower limbs are more associated with load carriage injury with up to a 15% increase in knee pain[65–68].

Fatigue

Effects on Lower Limb

Along with having to carry heavy body borne loads, soldiers are required to walk for prolonged periods of time, resulting in fatigue, which occurs when muscles fail to produce maximum muscular voluntary force[28,29]. During prolonged load carriage, mixed results have been shown for knee flexion/extension angle and range of motion[11,31,32,69]. However, peak vertical ground reaction force increases with the onset of fatigue, leading to an increase in knee flexion angle and impulse in an attempt to stabilize the knee joint[20,30,32,70]. These changes are thought to increase knee joint loading and overuse injury risk[30].

As fatigue increases, stride time and length decrease, and ground contact time and step width variability increases[31,58,69]. While these alterations help stabilize the body, these are factors that indicate a decrease in stability during muscular fatigue. However, hip range of motion increases in order to help absorb impact forces while fatigued[31]. These kinematic changes help stabilize the joints, but also increase muscle strain[31]. Hip internal rotations and moments increase, leading to an increase in knee abduction moment[30]. These increases show a decrease in muscle stabilization and an increase in medial compartment loading at the knee, ultimately increasing the risk of ACL injury[30]. Also, muscle function at the knee decreases after a prolonged period of time, which can increase the risk of musculoskeletal injury[58,71]. Energy cost and rate of perceived exertion also increase over time during prolonged load carriage, as seen by an increase in $VO_{2,max}$ and an increase in heart rate[11,33,72].

Combined with Load Carriage

When fatigue is combined with load carriage, such as in a prolonged load carriage task, these effects significantly increase, with detrimental effects being seen as early as 45 minutes into the task[73]. Peak vertical and sagittal ground reaction force increases, leading to an increase in knee flexion/extension range of motion and peak flexion angle[32,33]. Knee flexion impulse also increases from the additional forces placed on it[32]. However, despite this information about the effects of fatigue in the sagittal plane of motion, there is still limited data on the effects of fatigue on frontal plane knee motion.

Jerk and Jerk Cost

Joint smoothness may be an effective way to show joint instability and injury risk. Physiologically, joints produce the smoothest motion possible during the planning phase of gait[38]. Jerk cost is the accepted way to quantify joint smoothness, where a smooth joint trajectory will have a jerk cost close to zero[36–38,40,41,57]. Jerk cost originally was formed to predict joint smoothness of a single joint, but has been revised for multi-joint prediction[36]. During the initial stance phase, because the angular acceleration starts to decrease, angular knee jerk cost is high. Similarly, angular knee jerk cost is high during the last part of stance phase because an increase in angular acceleration occurs[41]. However, during execution, many factors affect whether the joint will be smooth.

Minimum Jerk Trajectory Model

Many studies regarding jerk aim to verify the validity of the minimum jerk trajectory model, which states that the smoothest motion is the one that will produce a jerk cost closest to zero[34,39,40,43]. Elite runners have a smoother gait than non-runners during running and walking tasks, showing a lower jerk cost in sagittal plane linear knee motion[38,70]. With practice, a joint produces smoother motion in all linear component directions[37,38]. Smooth and graceful movements can lead to better performance of rapid movements and more stability[35,37]. This smoothness is also translated into equipment that individuals use, with more skilled individuals having smoother motion with their equipment than less skilled[74]. Patients with knee osteoarthritis have also been seen to have a larger jerk cost in the knee than healthy subject[41].

Because higher peak jerk results in less stable movement, movement tasks are divided up into small tasks that favor stability over momentum[35]. However, a higher peak jerk for tasks such as lifting can suggest that the instability of the movement is compensated by stability in another part of the body and an increase in loading on the associated joint, leading to an increase in injury risk[34,35]. Quantifying smoothness of the joint can help quantify instability, but also injury risk in the body.

Inertial Measurement Units (IMUs)

Currently, most kinematic data are collected using a motion capture system. However, these systems can be expensive to install and only provide a limited capture volume. Inertial measurement units (IMUs), acceleration-based sensors, allow for unencumbered kinematic data analysis outside of a laboratory setting. These sensors are also a lot less expensive than a motion capture system and, because they constantly collect data, allow for a wider collection range than a motion capture system. However, before using IMUs as a replacement for a motion capture system, they need to be determined to be equivalent to the motion capture system, the gold standard of kinematic data collection.

Calculating Joint Kinematics

IMUs successfully calculated lower limb 3D kinematics, including 3D knee joint rotations[48,49,75]. Hip joint centers have also been able to be accurately calculated using IMUs[50]. IMUs have been determined to reliably gather repeatable acceleration data for a walking task, with the shank having the largest repeatability[46,76]. IMUs have been proven to be as good as motion capture systems in estimating joint kinematics in the sagittal region for a variety of tasks, including balancing, walking, and running[45,76]. While the IMUs are accurate, change in speed, for instance walking to running, creates some error during the transition period[76].

Along with testing the IMU devices, the accuracy of the algorithms that interpret the IMU data must also be verified. Also, a key to using IMUs is that the algorithms need to be resilient to variations in placement along the segment. Because IMUs are placed on rigid segments instead of bony anatomical landmarks, they are more prone to deviation in placement. Cooper et al. tested one algorithm's accuracy with deviations in placement and found that the algorithm was accurate for differing placements[76]. As IMUs become more popular, ease-of-use algorithms are being designed that accurately calculate lower limb 2D and 3D joint kinematics[51]. Similar to a motion capture system, skin motion artifact also plays a role in the error in kinematic calculations with IMUs[51]. The data from IMUs is usually calculated in the sagittal plane, but examining the data in the frontal plane of motion can help in identifying injury risk in a wide variety of environments.

<u>Validity in Determining Gait Parameters</u>

The validity and repeatability in using IMUs to determine gait parameters, such as stride speed, stride length, and walking detection is a well-researched topic. For spatiotemporal parameters, the closer the IMU is to the ground, such as on the foot, the more accurate the parameter, with accuracy similar to a motion capture system[77–81]. Additionally, IMUs are sensitive enough to detect small changes in gait, such as the speed of walking with eyes opened and closed, with both shank and foot sensors providing accurate speed measurements[78,82,83]. While walking speed has been reliably estimated, walking incline has shown differing results on the accuracy of calculation[80,82]. These parameters have been shown to be calculated accurately using IMUs for both healthy and neurologically impaired adults, whose gait is severely altered[84]. Trunk lean during walking has also been successfully examined using IMUs[85]. Overall, IMUs are becoming a reliable technology for calculating gait parameters.

CHAPTER THREE: MANUSCRIPT

Introduction

Musculoskeletal injuries are a common and costly military problem. Up to 12% of military personnel suffer a musculoskeletal injury each month, resulting in over 2.4 million health care visits annually, with a resultant cost of $700 million[1,2,86]. A majority of these musculoskeletal injuries are overuse and occur at or below the knee during training activities, with the most common injuries being stress fractures and sprains, which can create additional loading on the knee, potentially leading to knee osteoarthritis[1,4,5,87,88]. During training activities, soldiers routinely carry body borne loads between 20 kg and 40 kg, which reportedly increase musculoskeletal injury risk by altering lower limb neuromechanics[6–10,89]. Ground reaction forces in the vertical and antero-posterior plane increase, as well as knee flexion and adduction angles.

Military training may increase knee musculoskeletal injury risk[1,12,13]. During specific military training activities, such as prolonged walking, the addition of body borne load produces alterations of knee biomechanics thought to stabilize the joint. But, these alterations may also increase hazardous loading of joint's soft tissues and injury risk[14]. When walking with body borne load, vertical ground reaction forces reportedly increase between 12 and 50%[13–15,17–20]. To stabilize the limb in general and knee specifically, individuals reportedly increase knee flexion to help attenuate the elevated impact forces[7,13,14,18,32]. The increased knee flexion posture purportedly leads to larger joint contact forces, loading the knee's soft-tissue structures and increasing injury risk, which has been linked to radiographically confirmed knee osteoarthritis[13–15,18,20,87,90]. Moreover, walking with body borne load also increases both the angle and magnitude of the mediolateral ground reaction force[15,19,21]. A larger and more medially-directed ground reaction force acts to push the knee into varus and increases peak knee adduction angle and moment. Greater peak knee adduction joint angle and moment increase medial joint compartment loading and are related to the progression of knee injury and pain, as well as joint musculoskeletal disease (i.e., osteoarthritis)[23–27]. In particular, the external knee adduction moment is reportedly a correlate of medial knee joint compartment loading, and may decrease mediolateral stability of the joint[19,22,23]. Yet, despite the direct link to musculoskeletal injury, there is currently a dearth of information about frontal plane knee biomechanics (i.e., joint adduction angle and moment) during locomotion with body borne load.

Prolonged walking with body borne load may further increase changes to knee biomechanics during locomotion and increase knee musculoskeletal injury risk[28–30]. When walking for long periods of time without body borne load, individuals increase in vertical ground reaction force, resulting in an increase in peak knee flexion angle and impulse, with changes being observed at the end of the prolonged walking task[20,30–32]. Additionally, when walking without body borne load for 12.8 km, peak knee extensor moment significantly increases. With the addition of body borne load, knee biomechanics, such as peak knee flexion angle and moment, are reported to further increase after walking for 2 km, much earlier than with unloaded prolonged walking[20,31–33]. These biomechanical changes are thought to increase knee joint loading and overuse injury risk[30]. When walking with load for prolonged periods of time, anteroposterior and vertical ground reaction force increase, starting at as soon as 15 minutes after the beginning of walking, which can increase the risk of musculoskeletal injuries, such as stress fractures[20,32,33]. With the amount of foot strike impacts that occur during prolonged walking, these changes increase overuse injury risk[32]. However, to date there is limited information about the effects of prolonged load carriage in the frontal plane of knee motion.

Quantifying smoothness of knee motion using angular jerk, the rate of change of acceleration, may be an effective measure of injury risk[34,35]. In fact, angular jerk cost may be the best way to quantify the smoothness of joint movement, as the movement trajectory with the smoothest motion also exhibits the lowest jerk cost[36–40,43,44]. Quantitatively, jerk cost is:

$$Jerk\ Cost = \frac{1}{2}\int_0^t \left(\frac{d^3\theta}{dt^3}\right)^2 dt,$$

where t is time (sec) and θ is knee angle (rad)[36]. A movement trajectory that minimizes jerk uses less energy to execute and places smaller loads on the joint, reducing risk of injury[34,35]. Conversely, joint movements that have higher jerk are thought to have less coordination, which could lead to a higher risk of musculoskeletal injury, such as fracture[91]. In patients with radiographically confirmed musculoskeletal disease, angular jerk cost of the knee increases in both the sagittal and frontal planes of motion[41,42]. Additionally, in both healthy subjects and subjects with knee osteoarthritis, during the initial stance phase, because the angular acceleration is decreasing, angular knee jerk cost is high[41]. Similarly, angular knee jerk cost is high during the last part of stance phase because an increase in angular acceleration occurs[41]. While radiographically confirmed disease increases jerk, practice has been shown to decrease jerk, with runners showing lower jerk than non-runners, who are at more risk of sustaining an injury during running, in a running task[37,38]. Since experts and healthy subjects exhibit lower jerk, jerk is reportedly thought to increase with an increase in fatigue because fatigue has similar effects on motor control[92]. Additionally, in tasks such as lifting, peak jerk has been shown to increase with subjects who risk postural stability[91]. However, currently the effects of prolonged load carriage on angular knee jerk are unknown.

Traditionally, kinematic data are recorded with a motion capture system. However, motion capture systems are expensive, often limited to a laboratory setting, and have difficulty collecting data that represents day-to-day activities. Researchers have recently started using inertial measurement units (IMUs) to calculate kinematic data, as they are cheaper and can continuously collect acceleration-based data[45–48]. The main achievements for IMU-derived kinematic data have been to create algorithms to quantify sagittal plane joint kinematics, particularly at the knee[48–51]. Currently, however, IMU derived kinematics data are limited to sagittal plane, and it is relatively unknown how these measures compare to motion capture derived metrics. With that in mind, this study aimed to fill that critical void and test the effects of prolonged load carriage on angular jerk for frontal and sagittal plane knee motion with both IMUs and a motion capture system, and compare the equivalency between the IMU- and the motion capture-derived angular knee jerk. It is hypothesized that the addition of body borne load and increase in duration will significantly increase in peak and cost of angular jerk for sagittal and frontal plane knee motion. Additionally, it is hypothesized that the IMU- and motion capture-derived peak and cost of angular jerk will be statistically equivalent for both sagittal and frontal plane knee motion.

Methods

Participants

Eighteen participants (12 male, 6 female) were recruited for this study (Table 3.1). To be included, participants had to be healthy, recreationally active, and between 18 and 40 years old, as determined by a pre-participation and PAR-Q questionnaires (APPENDIX A, APPENDIX B)[93]. Each potential participant self-reported the ability to safely the carry up to 75 pounds (34 kg). Potential participants were excluded if they had: 1) history of back or lower extremity injury or surgery, 2) current back or lower extremity pain or injury, 3) known neurological disorder, or 4) were pregnant at the time of the study. Research approval from Boise State University's Institutional Review Board was obtained and each participant provided written consent prior to participation.

Table 3.1 Mean (SD) participant demographics.

	n	Height (m)	Weight (kg)	Age (years)
Males	12	1.81 (0.06)	77.91 (9.95)	23.33 (1.87)
Females	6	1.32 (0.08)	59.88 (3.79)	22.83 (4.79)

Experimental Design

Each participant completed one orientation and three test sessions. Each test session required the participant to complete the prolonged load carriage task with a different body borne load (0, 15, and 30 kg). For each load configuration, participants wore a spandex top and shorts. For the 15 and 30 kg configurations, an adjustable weighted vest (V-Max, WeightVest.com, Inc., Rexburg, ID, USA) was added to the participant's torso and systematically adjusted to provide the necessary load (Picture 3.1). Prior to testing, each participant was weighed to ensure the load was within ± 2% of the target weight. The load testing sequence was randomized prior to testing using a 3 x 3 Latin square (Table 3.2). Each test session was separated by a minimum of 24 hours to limit the effects of fatigue.

Picture 3.1 The spandex and weighted vest that participants wore for the 15 kg and 30 kg load conditions.

Table 3.2 The 3 x 3 Latin Square used for randomization of testing order for each weighted condition.

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

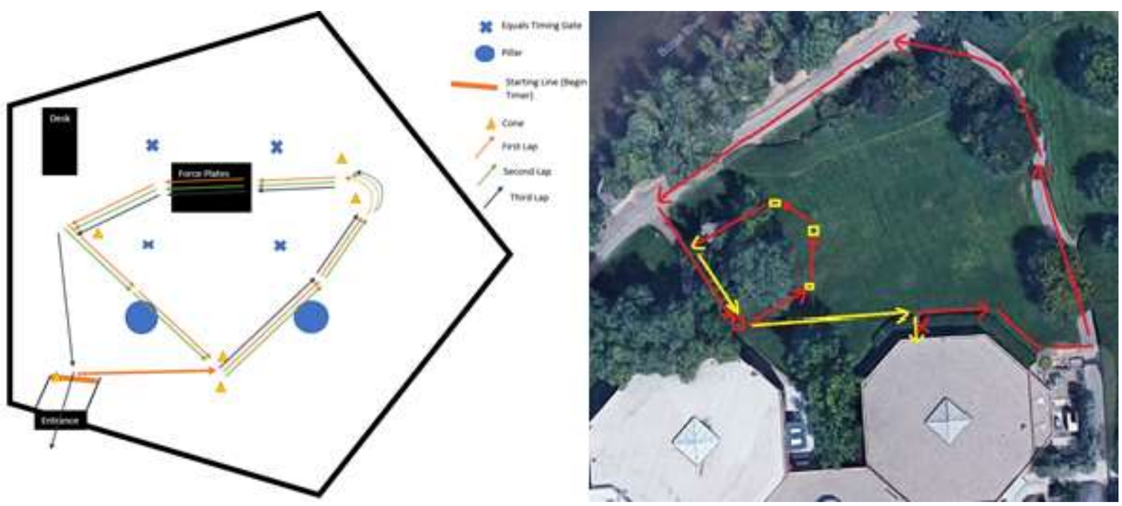
Orientation Session

Each participant completed one orientation session. During the orientation session, participant demographic (height, weight, age, and sex) and strength (trunk and lower limb) data were recorded, dominant limb was determined using the Waterloo Footedness Questionnaire (APPENDIX C)[94], and participants were familiarized with the study procedures. To record trunk strength, each participant performed a flexor endurance test, a modified Biering-Sorensen, and a side bridge test according to previous literature[95,96]. To record lower limb strength, each participant performed three maximum hip flexion/extension and abduction, knee flexion/extension, and ankle plantar-/dorsiflexion isometric contractions with the dominant limb on an isokinetic dynamometer (HUMAC, Computer Sports Medicine, Inc., Stoughton, MA). For hip flexion/extension participants stood with their hip flexed at 15-degrees[97]. For hip abduction, participants laid on their non-dominant side with their hip held in a neutral position (0-degrees)[98]. For knee flexion/extension, participants sat with their thigh secured and knee flexed at 60-degrees[97]. For ankle plantar-/dorsiflexion, participants laid prone with their ankle at 0-degrees[98]. Each contraction required participants to contract maximally for 3 seconds, and maximum torque was recorded. To conclude the orientation session, participants walked 1.3 m/s with each body borne load (15 and 30 kg) to ensure they were comfortable with the walking task and each load configuration.

Biomechanical Test Sessions

During each test session, participants had 3D lower limb (hip, knee, ankle) biomechanical data recorded while they walked over-ground for 60-minutes at 1.3 m/s. During the walk task, lower limb motion data was recorded with 8 high-speed optical cameras (240 Hz, MXF20, Vicon Motion Systems LTD, Oxford, UK) and 8 inertial measurement units (IMUs) (128 Hz, Opal, APDM, Inc., Portland, OR), while one ground embedded force platform (2400 Hz, OR6, Advanced Mechanical Technology, Inc., Watertown, MA) captured synchronous ground reaction force (GRF).

The walk task required participants complete 13 laps on an over-ground walking course in 60 minutes (Picture 3.2). The walking course was approximately 390 meters and composed of indoor and outdoor portions. Each participant started indoors at minute 0 and completed one lap (indoor plus outdoor) of the walking course every five minutes thereafter (minute 5, 10, 15, …, 60). For the indoor portion, participants walked 1.3 m/s ± 5% through the motion capture volume three times. During each walk trial, two sets of infrared timing gates (TracTronix TF100, TracTronix Wireless Timing Systems, Lenexa, KS), placed 4 m apart in the capture volume, recorded walking speed. Each walk trial was recorded as successful or unsuccessful. A successful trial required the participant to walk ± 5% of the target speed (1.3 m/s) and only contact the force platform with their dominant limb. After completing the indoor portion, participants completed the outdoor portion where they walked to a metronome (Planet Waves PW-MT-01, D’Addario, Farmingdale, NY) to ensure proper speed throughout the task.



Picture 3.2 The mapped-out walking course.

Biomechanical Analysis

During each trial, lower limb biomechanics were quantified from the 3D coordinates of 34 retro-reflective (15 mm diameter) and 4 virtual markers (Table 3.3), and 3D accelerations from 8 IMUs (Table 3.4). Each retro-reflective marker was placed on a bony landmark and secured using double-sided tape (Sensi-Tak Tape Roll, Walker Tape, West Jordan, UT) and elastic tape (Cover-Roll Stretch Tape, BSN Medical GmbH, Hamburg, Germany). Each virtual marker was digitized in the global coordinate system using a Davis Digitizing Pointer (C-Motion, Inc., Germantown, MD). The IMUs were placed on the participant and secured using a Velcro strap and elastic tape. After marker placement, each participant stood in anatomical position for a static recording. The static recording was used to create a kinematic model consisting of eight segments (trunk, pelvis, and bilateral thigh, shank, and foot) with 27 degrees of freedom in Visual 3D (v6, C-Motion, Inc., Germantown, MD). Each segment was assigned a local coordinate system and three orthogonal Cartesian axes. For the trunk, the origin was calculated as the intersection of the midpoint of the acromion processes and the midpoint of the C7 and clavicular notch, and assigned a local coordinate system with three degrees of freedom[99]. The pelvis was defined in relation to the global coordinate system, with the origin at the midpoint between the right and left iliac crests, and assigned six degrees of freedom (three rotational and three translational)[100]. For the hip, a functional joint center was calculated according to Schwartz and Rozumalski[101], and assigned a local coordinate system with three degrees of freedom. For the knee and ankle, the joint center was calculated as the midpoint between the medial and lateral epicondyles and malleoli, respectively, and assigned local coordinate systems with three degrees of freedom in accordance with previous literature[102,103]. After IMU placement, the participant performed a calibration routine to determine each sensors 3D relation and specific Cartesian axis. The calibration routine required each participant stand motionless in anatomical position for 10 seconds, perform four toe touches, walk 10 m, turn around, and walk 10 m back.

Table 3.3 Placement of the markers for the kinematic model

	Markers
Trunk	*xiphoid process*, *clavicular notch*, *C7 vertebrae*, *bottom of the scapula*, **acromion process**
Pelvis	**Anterior-Superior Iliac Spines**, **Posterior-Superior Iliac Spines**, Iliac Crests
Thigh	Greater Trochanter, **Lateral and Medial Femoral Epicondyles**, Distal Thigh
Shank	Tibial Tuberosity, Lateral Fibula, Distal Tibia, **Lateral and Medial Malleoli**
Foot	**First and Fifth Metatarsal Heads**, Heel, Midpoint between first and fifth metatarsal heads.

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

Table 3.4 Placement of the IMUs on the participant.

	IMU Placement
Upper Body	Sternum, Sacrum
Lower Body	Bilateral Thigh, Shank, Foot

For each walk trial, the 3D marker and IMU data were filtered through a fourth-degree low-pass Butterworth filter at 12 Hz and 15 Hz, respectively[104]. Then, the filtered marker data were processed in Visual 3D to calculate 3D knee rotations, which were expressed relative to the participants anatomical position using a joint coordinate system approach[102,103]. The filtered IMU data were processed with custom MATLAB (MATLAB r2018a, Mathworks, Natick, MA) code to calculate knee flexion-extension and abduction-adduction joint angles, similar to previous research[45,48,50]. Next, the first, second, and third derivates of knee flexion-extension and abduction-adduction joint rotations were calculated from the marker (Motion Capture) and IMU data to obtain angular velocity, acceleration, and jerk (Figure 3.1) with custom MATLAB code. Jerk cost was calculated as the sum of angular jerk for knee flexion-extension and abduction-adduction across stance phase according to:

$$Jerk\ Cost = \frac{1}{2}\int_0^t \left(\frac{d^3\theta}{dt^3}\right)^2 dt,$$

where t is time (sec) and θ is position (rad)[36]. All biomechanical data were normalized from 0% to 100% of stance phase and resampled to 1% increments (N = 101). Stance phase was defined as the time between initial contact and toe off, defined as the first instance the vertical ground reaction force exceeds and falls below 10 N, respectively.

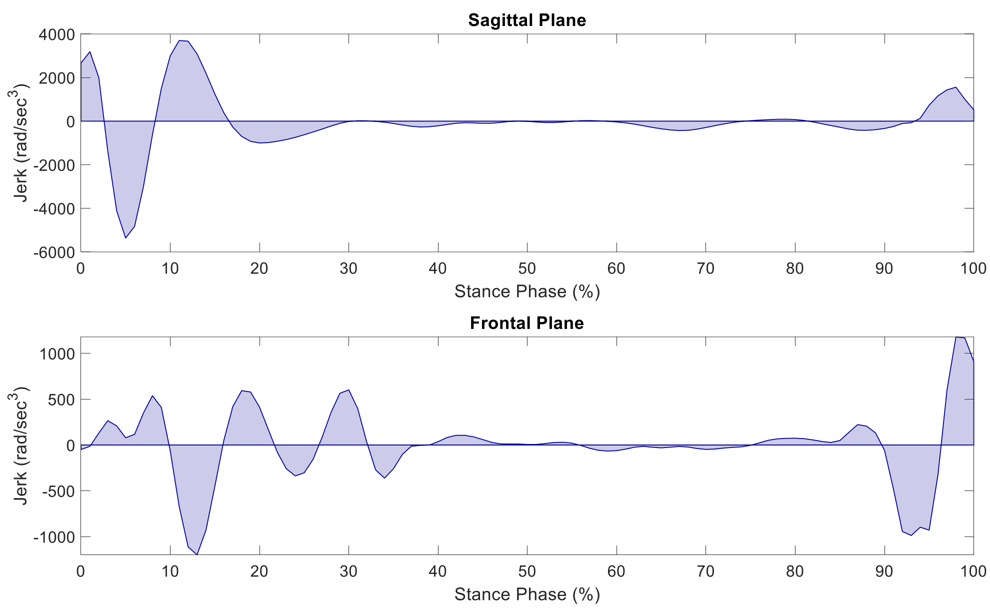


Figure 3.1 Jerk over time in the sagittal and frontal planes of knee motion.

<u>Statistical Analysis</u>

Biomechanical variables related to knee musculoskeletal injury risk were submitted to statistical analysis. The dependent variables were peak and cost of angular jerk for sagittal and frontal plane knee motion derived from both motion capture and IMU data. For each dependent variable, two successful trials from each time point (minutes 0, 5, 10 … 60) were averaged to create a participant-based mean. Prior to analysis, all dependent variables were checked for normal distribution using Shapiro-Wilk test[105], and all variables underwent a logarithmic transformation to achieve normality, in accordance with previous literature[41,42,106,107]. Then, the motion capture- and IMU-derived measures were submitted separately to a repeated measures linear mixed model with body borne load (0, 15, and 30 kg) and time (0, 15, 30, 45, and 60 min.) as fixed effects, and subject as random effects. A compound symmetry covariance structure was chosen to account for the correlation between each dependent variable with each load and at each time point. Significant interactions were submitted to simple main effects analysis, and a Hommel-Bonferroni correction was used for pairwise comparsions[108]. To compare motion capture- and IMU-derived measures of knee jerk with each load (0, 15, and 30 kg) and at each time point (0, 15, 30, 45, and 60 min), two one-sided tests (TOSTs) were performed with smallest effect size of interest ($d = 0.5$) and confidence interval of 90%, in accordance with Lakens[109,110]. Statistical analysis was run using SPSS (v25, IBM, Armonk, NY) and Excel (Microsoft, Redmond, WA) for the linear model and TOST analysis, respectively. Alpha was set a priori at $p < 0.05$.

Results

No significant interactions were observed. Thus, only significant main effects are presented below.

Motion Capture Derived Jerk

Body borne load significantly increased peak ($p < 0.001$) and cost ($p < 0.001$) of angular jerk of sagittal plane knee motion (Figure 3.2) (APPENDIX E). Specifically, there was a significant increase in peak and cost of sagittal plane knee jerk with the 30 compared to the 15 ($p < 0.001$, $p < 0.001$) and 0 kg ($p < 0.001$, $p < 0.001$) loads, but only jerk cost increased with the 15 compared to the 0 kg load ($p < 0.001$). Time had no significant effect on peak ($p = 0.351$) or cost ($p = 0.885$) of angular jerk of sagittal plane knee motion.

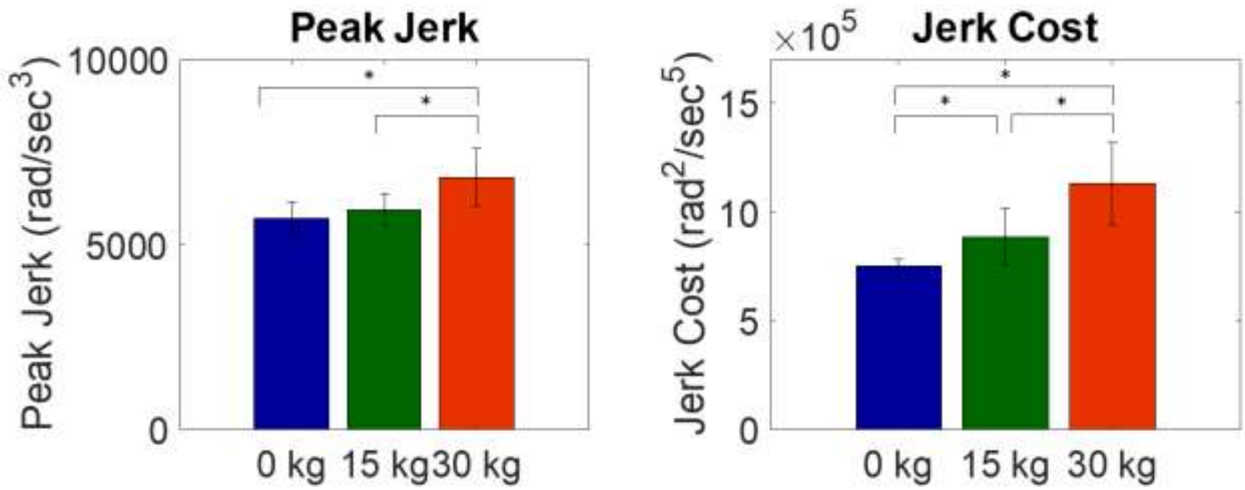


Figure 3.2 Peak (mean ± SD) and cost of angular jerk of sagittal plane knee motion with each body borne load. *Denotes significant difference ($p < 0.05$) between loads.

Body borne load significantly increased peak ($p < 0.001$) and cost ($p < 0.001$) of angular jerk of frontal plane knee motion (Figure 3.3). Specifically, there was a significant increase in peak and cost of frontal plane knee jerk with the 30 compared to the 15 ($p = 0.005$, $p = 0.001$) and the 0 kg ($p < 0.001$, $p < 0.001$) loads, and the 15 compared to the 0 kg load ($p = 0.001$, $p = 0.002$). Time had a significant effect on cost ($p = 0.001$), but not peak ($p = 0.084$) of frontal plane knee jerk (Figure 3.4). Frontal plane jerk cost increased at minute 60 compared to minutes 15 ($p = 0.004$) and 0 ($p < 0.001$), and at minute 45 compared to minute 0 ($p = 0.004$). Significant differences were not observed between any other time points ($p > 0.05$).

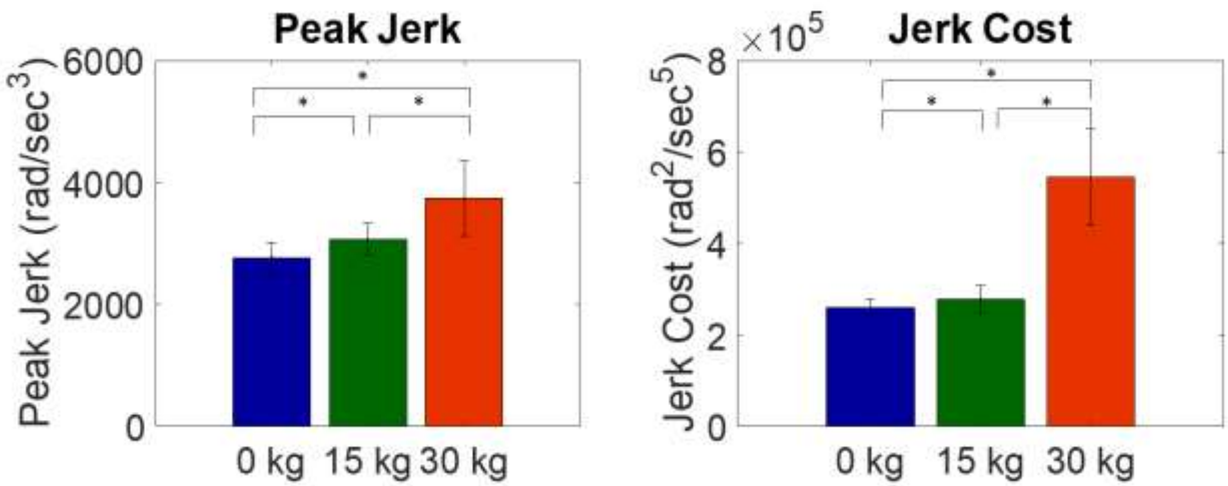


Figure 3.3 Peak (mean ± SD) and cost of angular jerk of frontal plane knee motion with each body borne load. *Denotes significant difference ($p < 0.05$) between loads.

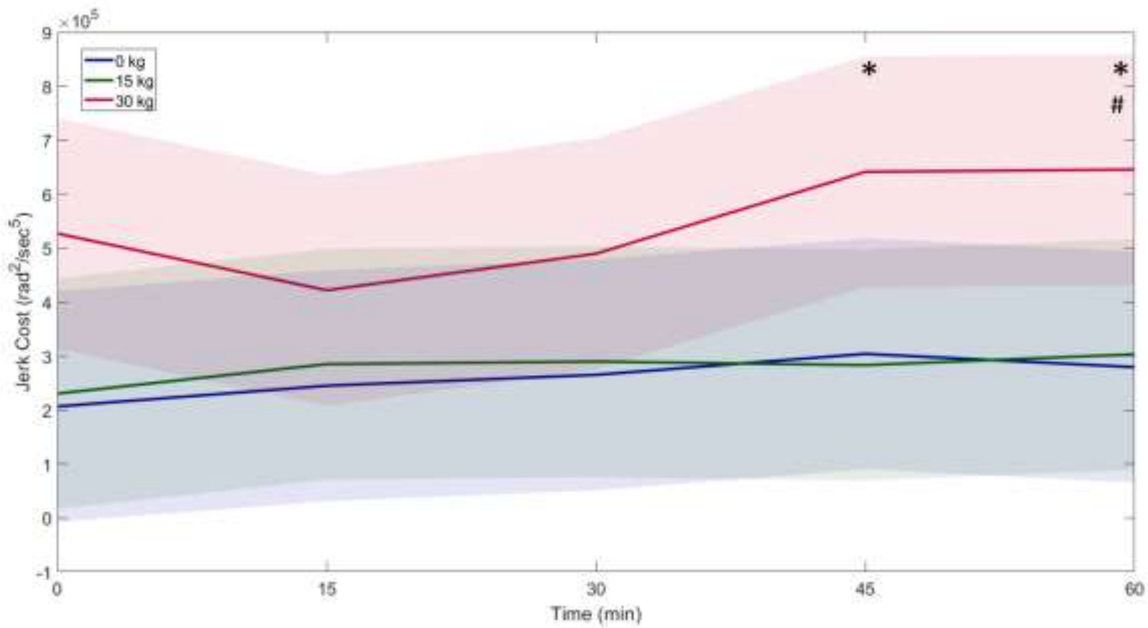


Figure 3.4 Jerk cost of frontal plane knee motion over the duration of the prolonged load carriage task. *,# Denote significant difference ($p < 0.05$) compared to minute 0 and 15, respectively.

<u>IMU Derived Jerk</u>

Body borne load significantly increased peak ($p < 0.001$) and cost ($p < 0.001$) of angular jerk of sagittal plane knee motion (Figure 3.5) (APPENDIX E). There was a significant increase in peak and cost of sagittal plane knee jerk with the 30 ($p < 0.001$, p

< 0.001) and 15 (p < 0.001, p = 0.044) compared to the 0 kg load. But, only jerk cost increased with the 30 compared to the 15 (p = 0.002) and the 15 compared to the 0 kg (p = 0.024) load. Time had no significant effect on peak (p = 0.987) or cost (p = 0.936) of angular jerk of sagittal plane knee motion.

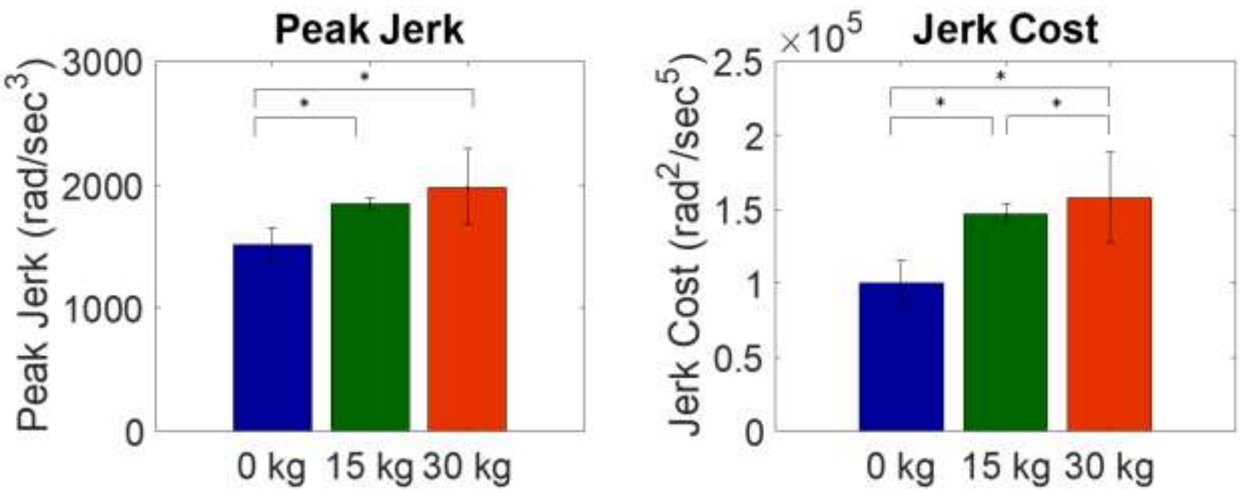


Figure 3.5 Peak (mean ± SD) and cost of angular jerk of sagittal plane knee motion with each body borne load. *Denotes significant difference (p < 0.05) between loads.

Body borne load significantly increased peak (p = 0.027) and cost (p < 0.001) of angular jerk of frontal plane knee motion (Figure 3.6). Specifically, there was a significant increase in peak and cost of frontal plane knee jerk with the 30 compared to the 0 kg (p = 0.010, p < 0.001) load, while only jerk cost increased with the 30 compared to the 15 kg load (p < 0.001). Time had a significant effect on cost (p = 0.015), but not peak (p = 0.158) of frontal plane knee jerk (Figure 3.7). There was a significant increase in jerk cost at minute 60 compared to minutes 15 (p = 0.004) and 0 (p = 0.003), but significant differences were not observed between any other time points (p > 0.05).

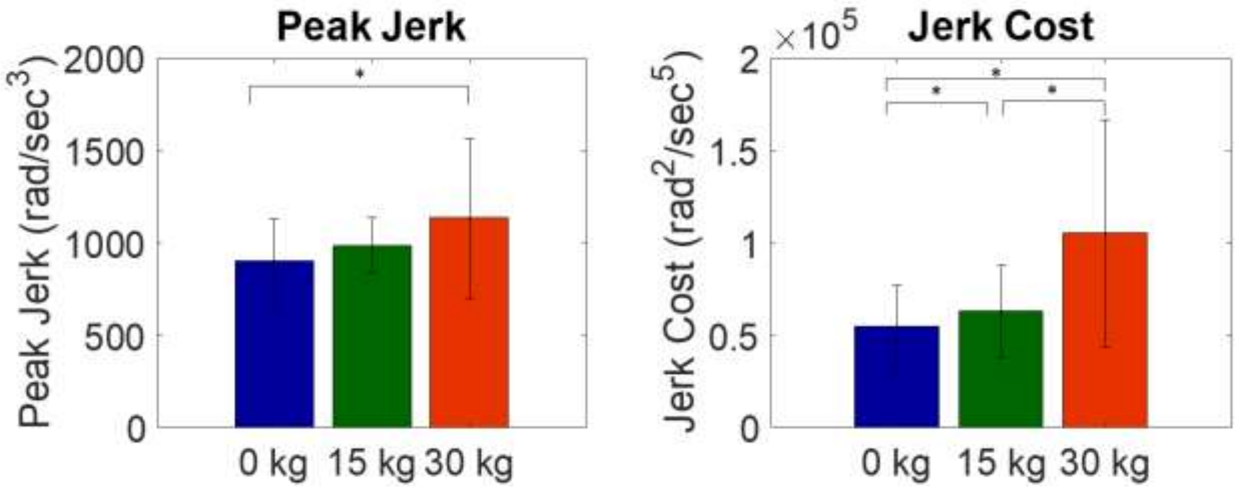


Figure 3.6 Peak (mean ± SD) and cost of angular jerk of frontal plane knee motion with each body borne load. *Denotes significant difference ($p < 0.05$) between loads.

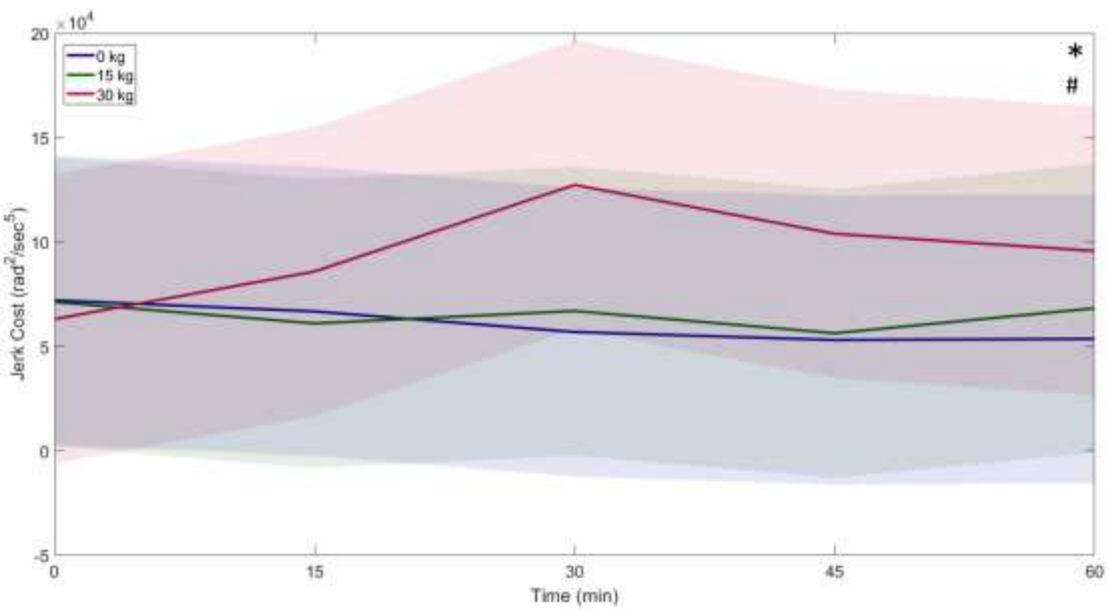


Figure 3.7 Jerk Cost of frontal plane knee motion over the duration of the prolonged load carriage task. *,# Denote significant difference ($p < 0.05$) compared to minute 0 and 15, respectively.

<u>Equivalence Tests</u>

The motion capture and IMU-derived measures of peak and cost of angular jerk for both sagittal and frontal planes of knee motion were not statistically equivalent for any body borne load ($p > 0.05$) or time point ($p > 0.05$) (Figure 3.8, Figure 3.9, Figure 3.10) (APPENDIX F).

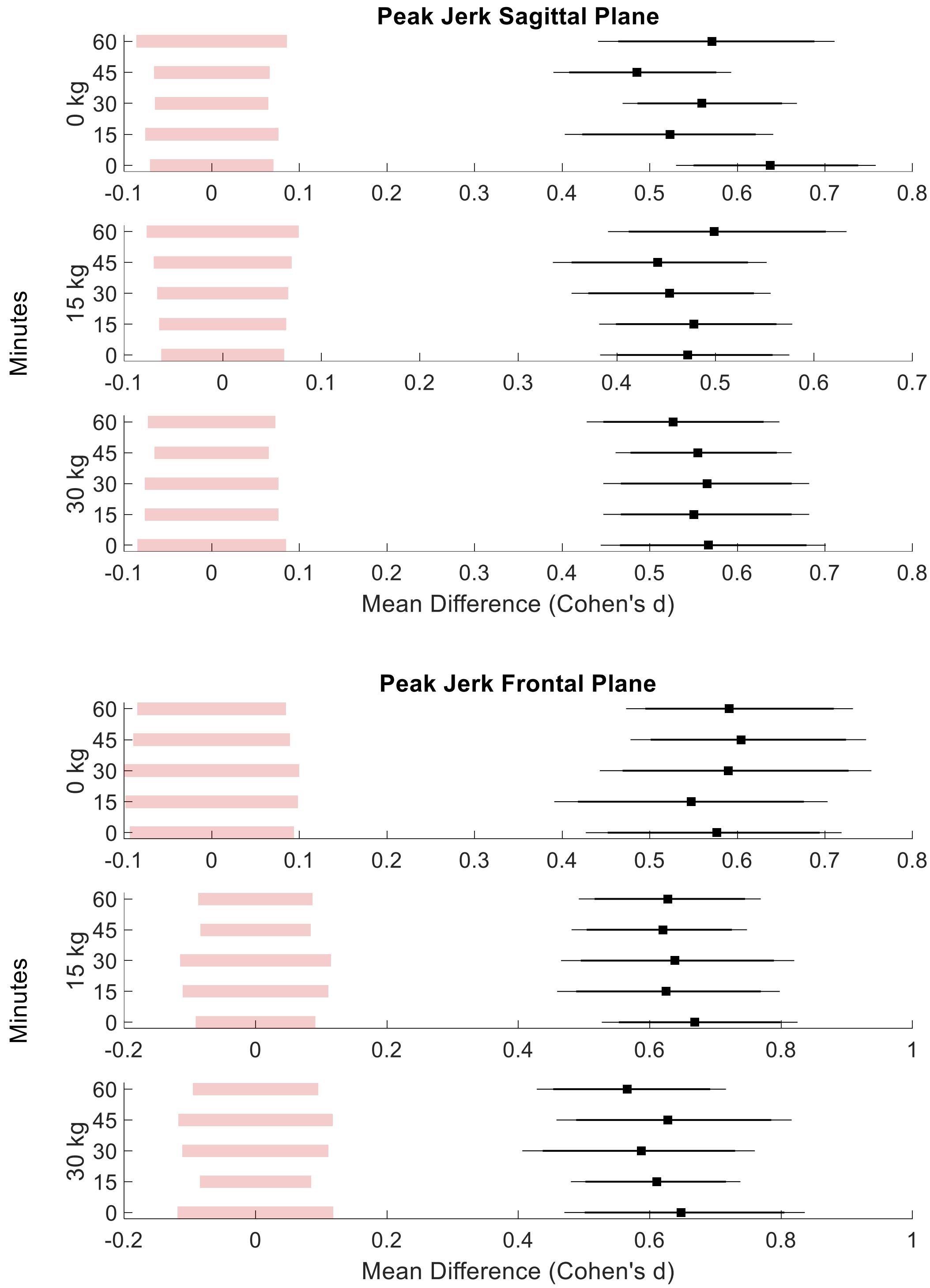


Figure 3.8 The mean difference between the motion capture and the IMU peak jerk (black square) and the 90% confidence interval (black line) compared to the equivalence bounds (red rectangle).

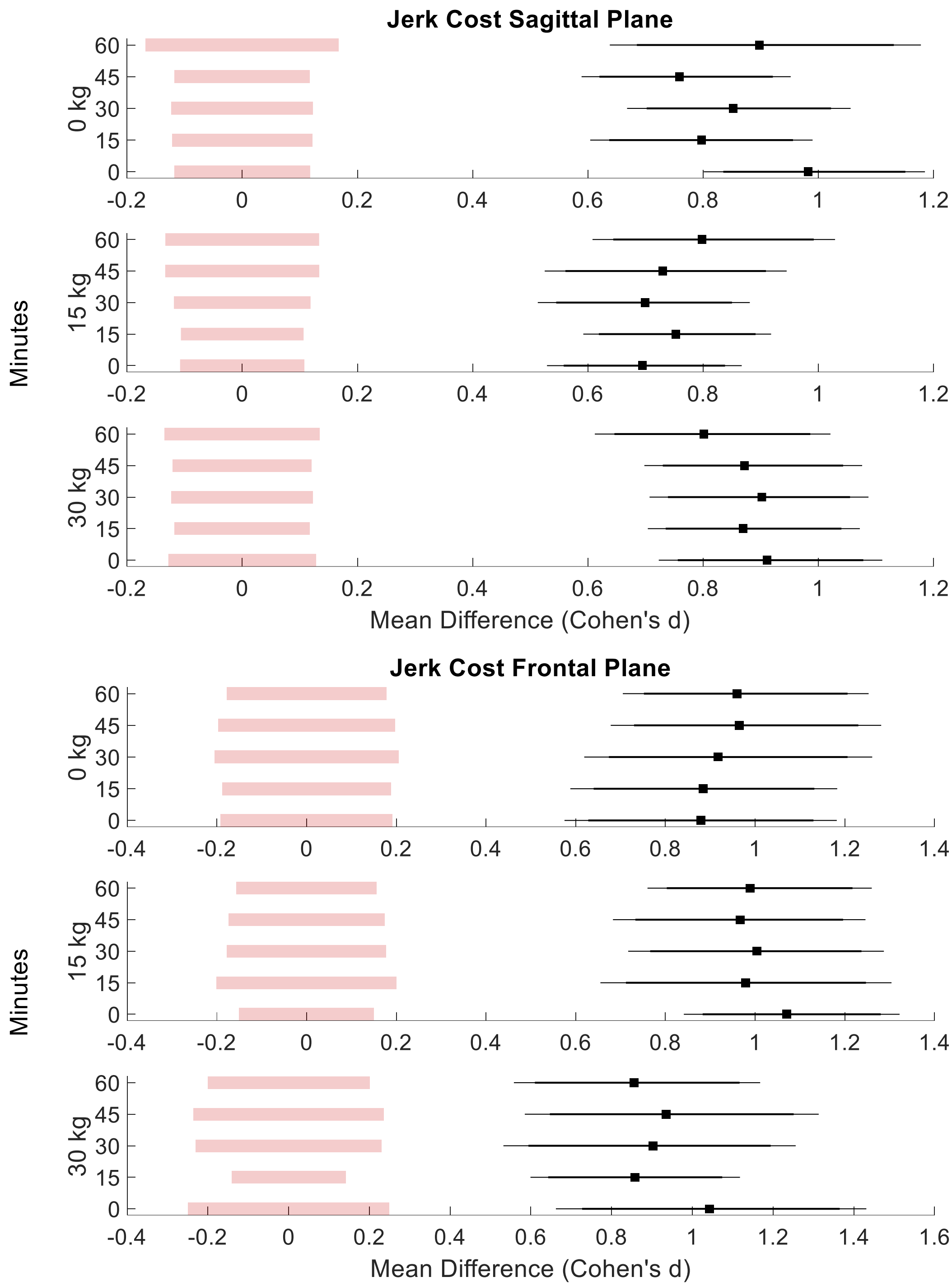


Figure 3.9 The mean difference between the motion capture and the IMU jerk cost (black square) and the 90% confidence interval (black line) compared to the equivalence bounds (red rectangle).

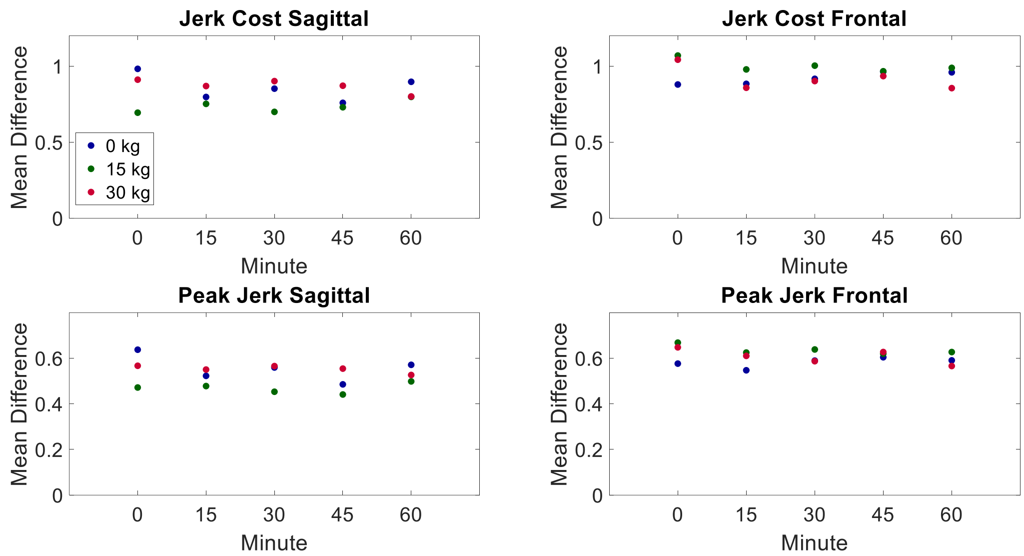


Figure 3.10 The mean difference between the motion capture-derived data and the IMU-derived data at each time point for each load condition.

Discussion

We sought to determine whether body borne load (0, 15, and 30 kg) and duration of walking (60 min) would increase jerkiness of both sagittal and frontal plane knee motions, and whether IMUs could accurately quantify those jerky knee motions. In partial support of our hypotheses, both body borne load and time increased jerky knee motions, but the IMU-derived jerk measures were not statistically equivalent to motion capture-derived measures.

Body borne load may increase jerky knee motion and risk of musculoskeletal injury. In support of our hypothesis, participants exhibited a significant 20% and 51% increase in peak and cost of knee sagittal plane angular jerk with the addition of body borne load. Jerkier knee motion decreases the smoothness of movement, which reportedly increases joint loading and injury risk[107,111]. During walking, for instance, the largest magnitudes of angular jerk cost occur during the weight acceptance phase, or early stance phase, when vertical ground reaction forces and lower limb joint torques are the highest[107]. Body borne load increases peak vertical ground reaction forces and lower limb joint torques, particularly at the knee, upwards of 10% and 98%, respectively[59,112]. These elevated ground reaction forces and knee joint torques result in greater response of lower limb musculature in general and associated knee musculature specifically[113,114]. This increased musculature activation places greater loads on the knee's soft-tissue structures, which, combined with the increased ground reaction forces, may subsequently increase musculoskeletal injury risk[13,15,17–19,90,107,113,115,116]. But, the larger ground reaction forces and joint torques may also compromise the individual's ability produce smooth motions, translating to the jerkier motions that are currently evident when walking body borne load.

In support of our hypothesis, the current participants also exhibited a significant 35% and 110% increase in peak and cost of knee frontal plane angular jerk with the addition of body borne load. Large frontal plane knee biomechanics, including significant increases in peak knee adduction angle and moment, are reportedly implicated in pathogenesis and progression of knee osteoarthritis[24,27,90,117]. The jerky frontal plane knee motions exhibited with the 15 and 30 kg body borne loads may accelerate the wearing of the joint's articular surfaces and increase likelihood of knee osteoarthritis development[41,42,106]. In fact, individuals with radiographically confirmed knee osteoarthritis exhibited 54% greater frontal plane angular knee jerk cost than healthy controls[42]. Considering the increase in frontal plane jerk cost with body borne load was more than double the increase in jerk cost of knee osteoarthritis patients compared to healthy individuals (110% vs 54%), walking with these heavy military-relevant body borne loads may lead to a substantial increase in risk of knee osteoarthritis development for service members. To decrease the risk of knee osteoarthritis development during service, the military may need to reduce jerkiness of knee motions, particularly in the frontal plane, during training-related activities. Furthermore, skilled runners exhibit smoother gait (i.e., decreased jerk cost of heel motion) than non-skilled runners[38]; future study is warranted to determine whether experienced load carriers exhibit smoother knee motions, and whether targeted military training programs can reduce jerkiness of knee motion and subsequent injury risk of inexperienced load carriers. Also, future study is warranted to determine the specific increase in jerk cost that elevates risk of musculoskeletal injury at the knee.

The duration of walking increased jerky frontal plane, but not sagittal plane, knee motion. In partial support of our hypothesis, participants exhibited a significant 31% increase in frontal plane jerk cost after 45 minutes of walking with body borne load. During a similar prolonged load carriage task, detrimental changes in gait, such as a significant increases in peak vertical ground reaction force, were reported to start after 15 minutes of walking[33]. These increases in vertical ground reaction force may further load the knee joint and present as jerkier motion, particularly in the frontal plane, as the duration of walking progresses. Larger vertical ground reaction forces and jerkier frontal plane knee motions may load the medial knee joint compartment, increasing the pain and loss of articular cartilage that characterize knee osteoarthritis[118–121]. Yet, significant increases in jerky knee motion were limited to the frontal plane, despite the fact that individuals reportedly exhibit greater peak sagittal plane knee angles and moments as duration of walking progresses[32,58]. The musculature responsible for sagittal plane knee motion have larger moment arms and can produce greater muscular force than the musculature responsible for frontal plane knee motion[122,123]. This enhanced muscular function may afford the individual greater resistance and attenuation of the elevated ground reaction forces, in addition to greater neuromuscular control in the sagittal plane, that result in smoother knee kinematics as duration of load carriage progresses[24,117]. As such, further study is warranted to determine whether increasing the strength of the frontal plane knee musculature can reduce jerkiness of those knee motions.

The IMU-derived measures of knee jerk also increased with the addition of body borne load and duration of walking. Parallel to the motion capture derived measures, the IMU derived peak and cost of sagittal and frontal plane knee angular jerk increased up to 77% with the addition of body borne load and up to 48% throughout the walking task. Although the IMU-derived measures of knee jerk had similar statistically significant increases with load and duration of walking as motion capture-derived measures, they were not statistically equivalent. In fact, contrary to our hypothesis, none of the currently tested IMU derived measures of knee jerk were statistically equivalent to the motion capture-derived measures. Although IMUs are reported to accurately calculate both the sagittal and frontal plane angles of knee motion[48,75], this did not translate to the current jerk calculations. It may be that the jerk calculations are sensitive to noise and/or drift that results from technical limitations of current IMU sensor technology. Currently, IMU sensors are limited to a sampling frequency of 128 Hz, which may not accurately record peak accelerations and may result in potential drift. To compensate for this potential drift, existing processing algorithms for joint rotations reset the joint (i.e., knee) angle to zero during the stance phase of each stride. However, because jerk is the third derivative of knee angle (position), and small discrepancies exist between recorded and “true” acceleration, whether the result of drift and/or sensor noise, calculated knee jerk measures may be inaccurate (APPENDIX D). Regardless, with technological advances in IMU sensors, further testing is warranted with higher frequency sensors to refine algorithms and accurately calculate knee jerk.

As mentioned above, the current study is limited by the sampling frequency of the IMU sensors. Jerk is a sensitive measurement, and an inadequate sampling frequency may lead to inaccurate recording of peak and/or directional changes in acceleration. Another potential limitation of the study is the chosen participants, who were not required to have load carriage experience. Considering, jerk is reported to differ by experience (or skill level)[38], replicating the current work with experienced load carriers is warranted. Many military recruits, however, enter training with minimal load carriage experience, and the results contained herein directly contribute to reduction of their injury risk during military related activities.

Conclusion

In conclusion, prolonged load carriage led to jerkier knee motion and increased knee musculoskeletal injury risk. Specifically, the addition of body borne load produced significant increases in angular jerk for both sagittal and frontal plane knee motion. These jerkier knee motions may increase loading at the joint, thereby increasing knee musculoskeletal injury risk. The duration of walking, however, only increased jerk for frontal plane knee motion, and individuals may be at greater risk of musculoskeletal injuries related to altered frontal plane knee motions when walking for long periods of time. Although the IMU-derived measures quantified similar increases in knee jerk as the motion capture-derived measures, the knee jerk values calculated by the two methods were not statistically equivalent.

CHAPTER FOUR: CONCLUSION

Introduction

This study's purpose was two-fold, (1) to examine the influence of prolonged load carriage on peak and cost of angular jerk of sagittal and frontal plane knee motion, and (2) to determine whether inertial measurement units (IMUs) can accurately calculate angular knee jerk. Key findings support the hypothesis that body borne load increases peak and cost of angular jerk in both the sagittal and frontal planes of knee motion, and duration of walking increases cost of angular jerk in the frontal plane knee motion. But, contrary to our hypothesis, the IMU-derived measures of jerk were not statistically equivalent to the motion capture-derived measures.

Key Findings

Prolonged load carriage led to jerkier knee motions and increased knee musculoskeletal injury risk. Specifically, the addition of body borne load resulted in a significant increase in peak and cost of angular jerk for both sagittal and frontal plane knee motion, while duration of walking increased frontal plane knee motion jerk cost. These jerkier knee motions may result from the musculoskeletal system's inability to adequately attenuate the elevated ground reaction forces and joint torques evident during load carriage, placing larger loads on the knee joint and increasing risk of musculoskeletal injury – particularly when walking for long periods of time. Additionally, the IMU-derived measures of knee jerk exhibited similar increases with the addition of body borne load and duration of walking as the motion capture-derived measures. However, the IMU-derived values of knee jerk were not statistically equivalent to the motion capture-derived values.

Significance

These findings support the tenet that prolonged load carriage resulted in detrimental changes to smoothness of lower limb motion that may increase knee musculoskeletal injury risk. This study is the first to document significant increases in peak and cost of jerk of both sagittal and frontal plane knee motion with the addition of body borne load and walking duration. Explicit kinematic changes may increase the loads place on the knee’s soft-tissue structures and elevate musculoskeletal injury risk. These experimental findings can be implemented by the military to reduce and monitor knee musculoskeletal injury risk during service. Specifically, the military can use this information to identify high-risk service members and quantify detrimental changes in knee biomechanics during military activities. This study also documented that IMU-derived measures of sagittal and frontal plane knee jerk detected similar statistically significant increases with the addition of body borne load and duration of walking as the motion capture-derived measures, but they were not statistically equivalent. As such, IMU technology may need further development to feasibly replace the motion capture system for accurately quantifying certain lower limb kinematics measures.

Limitations

This study may be limited by the IMU sensor sampling frequency. The IMU sensors used in this study have a sampling frequency of 128 Hz, which may be insufficient to accurately record linear and angular acceleration during dynamic movement tasks, such as walking with body borne load. Although this technological limitation may have impacted the current IMU-derived jerk measures, these sensors have previously been used to accurately calculate knee joint rotations. The study may also be limited by the chosen participants and load carriage configurations. The current participants were not required to have any load carriage experience and may exhibit different lower limb biomechanics than an experienced load carrier. For example, jerky gait motions are reportedly smaller for experienced (i.e., skilled) than inexperienced (i.e., unskilled) individuals during running[38]. Testing experienced load carriers might have produced different results. Nonetheless, most military recruits enter service with minimal load carriage experience, and limiting their rate of musculoskeletal injury has substantial physical and economic benefits. Additionally, the body borne load was currently applied to a participant's torso via a weighted vest, which may not be operationally relevant. Although the chosen loads accurately represent the weight service members are oftentimes required to carry during operational and training activities[6,89], a weighted vest may not accurately represent the load carriage equipment service members use during those activities. Regardless of the equipment used to carry the body borne load, the impact on knee biomechanics should not statistically differ[64].

Future Work

Knee biomechanics may differ between experienced and inexperienced load carriers – particularly with heavy military-relevant body borne loads. As such, future research is warranted to determine whether inexperienced load carriers exhibit jerkier knee motions, and whether targeted training programs can reduce these hazardous knee motions and musculoskeletal injury risk. Moreover, females reportedly exhibit a greater rate of knee musculoskeletal injury than males during military service[124], but it is unknown if they also exhibit jerkier knee motions during load carriage than males. Determining whether females present jerkier knee motions during military activities may provide avenue to reduce their injury risk. Considering knee jerk is 54% larger for knee OA patients than heathy controls[42], future study is warranted to determine the specific increase in jerk cost that elevates risk of knee musculoskeletal injuries, such ligament rupture or meniscal tear.

Inevitably IMU technology will improve such that the sensors will possess sampling frequencies that can accurately record acceleration during dynamic tasks. Future research is needed to determine the IMUs ability to record lower limb biomechanics, particularly knee jerk, during ecologically valid military settings.

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APPENDIX A

Pre-Participation Questionnaire

Pre-participation Questionnaire

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

YES NO

If yes, please describe: ____

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

YES NO

If yes, please describe: ____

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

YES NO

If yes, please describe: ____

4. Are you currently experiencing knee pain?

YES NO

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

YES NO

If yes, please describe: ____

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

YES NO

If yes, please explain: ____

I certify that the information I provided above is accurate.

Subject's Signature: ___________________________ Date: ______________

Subject's Name (Print): _________________________

Parent/Legal Guardian Signature: ___________________ Date: ___________

Parent/Legal Guardian Name (Print): ________________________

APPENDIX B

Participant Activity Rating Questionnaire

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

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

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

Scoring Criteria:
Low: PAS < 400
Moderate: 400 ≤ PAS < 560
High: PAS ≥ 560

APPENDIX C

Footedness Questionnaire

Footedness Questionnaire

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

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

Yes **No**

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

Yes **No**

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

APPENDIX D

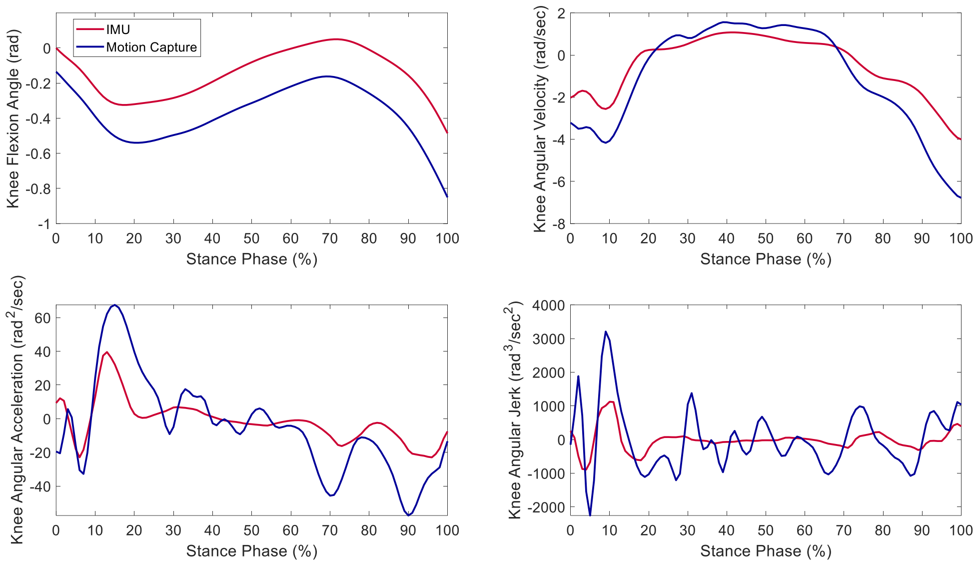

Figure D.1 An example of the effect of the difference in starting knee flexion angle between the motion capture angle (blue) and the IMU angle (red) on velocity, acceleration, and jerk.

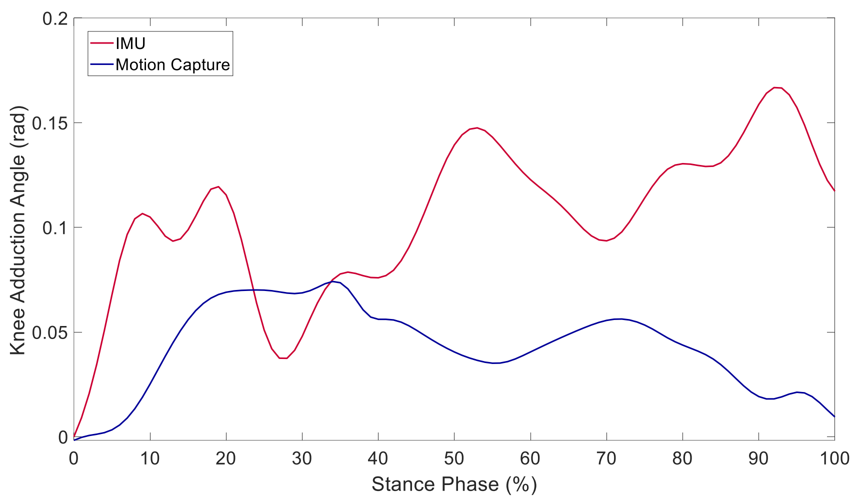

Figure D.2 An example of the correct motion capture-calculated frontal plane knee angle (blue) and the incorrect IMU-calculated frontal plane knee angle (red

APPENDIX E

Table E.1 **Mean (SD) motion capture-derived peak (x10^3 rad/sec^3) and cost (x10^5 rad^2/sec^4) of angular jerk.**

	0 kg					15 kg					30 kg				
	0 min	15 min	30 min	45 min	60 min	0 min	15 min	30 min	45 min	60 min	0 min	15 min	30 min	45 min	60 min
Peak Jerk Sagittal Plane[a]	6.22 (1.33)	5.69 (1.54)	5.85 (1.53)	5.36 (1.59)	5.64 (1.77)	6.10 (1.61)	6.03 (1.76)	5.83 (1.64)	5.96 (1.72)	6.10 (1.63)	7.40 (3.36)	6.54 (1.69)	6.87 (2.19)	7.10 (1.93)	6.65 (2.28)
Jerk Cost Sagittal Plane[a]	7.88 (2.75)	7.52 (3.14)	7.57 (3.44)	7.10 (4.24)	7.85 (4.47)	8.51 (3.31)	8.81 (4.03)	8.64 (3.52)	9.63 (4.60)	9.39 (3.96)	12.15 (8.53)	10.08 (4.44)	11.42 (5.65)	12.06 (5.65)	11.58 (6.88)
Peak Jerk Frontal Plane[a,b]	2.07 (1.78)	2.45 (1.65)	2.73 (2.15)	3.12 (2.89)	2.86 (2.26)	2.33 (1.12)	2.90 (1.97)	2.92 (1.78)	2.88 (1.39)	3.11 (1.59)	5.44 (8.08)	428 (4.43)	4.92 (4.57)	6.61 (7.02)	6.54 (6.43)
Jerk Cost Frontal Plane[a]	2.46 (0.87)	2.77 (0.98)	2.84 (1.17)	3.02 (1.47)	2.85 (1.13)	2.89 (0.86)	3.17 (1.17)	3.10 (0.90)	3.09 (0.85)	3.20 (0.98)	3.70 (2.72)	3.37 (1.56)	3.53 (1.40)	4.24 (3.03)	3.95 (1.88)

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

Table E.2 Mean (SD) IMU-derived peak (x10^3 rad/sec^3) and cost (x10^5 rad^2/sec^4) of angular jerk.

	0 kg					15 kg					30 kg				
	0 min	15 min	30 min	45 min	60 min	0 min	15 min	30 min	45 min	60 min	0 min	15 min	30 min	45 min	60 min
Peak Jerk Sagittal Plane[a]	1.52 (0.61)	1.56 (0.57)	1.46 (0.38)	1.46 (0.46)	1.45 (0.50)	1.84 (0.51)	1.90 (0.60)	1.83 (0.52)	1.87 (0.59)	1.83 (0.72)	1.89 (0.43)	1.84 (0.53)	2.02 (0.62)	1.88 (0.52)	2.02 (0.43)
Jerk Cost Sagittal Plane[a]	0.95 (0.67)	1.03 (0.68)	0.94 (0.56)	0.93 (0.54)	1.00 (0.80)	1.42 (0.75)	1.46 (0.86)	1.43 (0.75)	1.51 (0.98)	1.49 (0.50)	1.38 (0.50)	1.38 (0.85)	1.64 (0.81)	1.54 (0.89)	1.69 (0.70)
Peak Jerk Frontal Plane[a,b]	2.07 (1.78)	2.45 (1.65)	2.73 (2.15)	3.12 (2.89)	2.86 (2.26)	2.33 (1.12)	2.90 (1.97)	2.92 (1.78)	2.88 (1.39)	3.11 (1.59)	5.44 (8.08)	428 (4.43)	4.92 (4.57)	6.61 (7.02)	6.54 (6.43)
Jerk Cost Frontal Plane[a]	0.72 (0.88)	0.67 (0.91)	0.57 (0.67)	0.53 (0.37)	0.54 (0.41)	0.71 (0.81)	0.61 (0.64)	0.67 (0.66)	0.56 (0.48)	0.68 (0.81)	0.63 (0.68)	0.86 (1.11)	1.27 (1.90)	1.04 (1.13)	0.96 (1.13)

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

APPENDIX F

TOST Equivalence Test Results

Table F.1 The t values and p values for the TOST equivalence tests for sagittal plane jerk cost

					Equivalence Test		NHST Test	
Load Condition	Time Point (min)	MoCap Mean	IMU Mean	Degrees of Freedom	t-value	p-value	t-value	p-value
0 kg	0	5.8823	4.8894	16.82	9.635	1.000	10.935	< 0.001
	15	5.839	5.0421	20.41	7.29	1.000	8.604	< 0.001
	30	5.8381	4.976	21.43	7.929	1.000	9.247	< 0.001
	45	5.8035	5.0334	24.43	7.414	1.000	8.745	< 0.001
	60	5.8403	4.9322	18.02	5.462	1.000	7.067	< 0.001
15 kg	0	5.9027	5.2049	22.79	7.235	1.000	8.559	< 0.001
	15	5.8992	5.1442	25.91	8.155	1.000	9.495	< 0.001
	30	5.8854	5.1883	24.12	6.49	1.000	7.82	< 0.001
	45	5.9297	5.1948	21.98	5.94	1.000	7.261	< 0.001
	60	5.9389	5.1205	21.49	6.765	1.000	8.083	< 0.001
30 kg	0	5.9981	5.0811	26.98	8.337	1.000	9.693	< 0.001
	15	5.9721	5.0843	22.02	8.671	1.000	9.992	< 0.001
	30	6.0036	5.1066	24.5	8.385	1.000	9.717	< 0.001
	45	6.0477	5.1608	22.51	8.396	1.000	9.719	< 0.001
	60	5.995	5.1789	26.93	6.837	1.000	8.188	< 0.001

Table F.2 The t values and p values for the TOST equivalence tests for frontal plane jerk cost.

Load Condition	Time Point (min)	MoCap Mean	IMU Mean	Degrees of Freedom	Equivalence Test		NHST Test	
					t-value	p-value	t-value	p-value
0 kg	0	5.2108	4.332	20.49	4.715	1.000	6.03	< 0.001
	15	5.2977	4.4121	20.84	4.879	1.000	6.195	< 0.001
	30	5.321	4.3813	22.58	4.74	1.000	6.063	< 0.001
	45	5.3408	4.361	26.19	5.334	1.000	6.676	< 0.001
	60	5.348	4.3689	25.21	6.019	1.000	7.355	< 0.001
15 kg	0	5.3278	4.2463	18.81	8.097	1.000	9.405	< 0.001
	15	5.3802	4.4005	17.54	5.057	1.000	6.36	< 0.001
	30	5.3972	4.3955	18.38	6.067	1.000	7.373	< 0.001
	45	5.4	4.4355	17.42	5.924	1.000	7.226	< 0.001
	60	5.4502	4.4402	18.54	7.148	1.000	8.455	< 0.001
30 kg	0	5.4653	4.419	24.92	4.27	1.000	5.604	< 0.001
	15	5.4743	4.6154	26.97	5.435	1.000	6.792	< 0.001
	30	5.5352	4.6413	21.53	3.804	0.999	5.123	< 0.001
	45	5.6124	4.6632	25.18	4.036	1.000	5.371	< 0.001
	60	5.652	4.7889	26.82	4.453	1.000	5.801	< 0.001

Table F.3 The t values and p values for the TOST equivalence tests for sagittal plane peak jerk.

Load Condition	Time Point (min)	MoCap Mean	IMU Mean	Degrees of Freedom	Equivalence Test		NHST Test	
					t-value	p-value	t-value	p-value
0 kg	0	3.7912	3.1469	17.31	10.596	1.000	11.897	< 0.001
	15	3.7375	3.2155	22.42	7.757	1.000	9.08	< 0.001
	30	3.7511	3.1823	25.25	10.449	1.000	11.785	< 0.001
	45	3.7155	3.2237	25.24	8.655	1.000	9.99	< 0.001
	60	3.7311	3.1551	21.76	7.53	1.000	8.85	< 0.001
15 kg	0	3.7751	3.296	26.00	8.959	1.000	10.299	< 0.001
	15	3.762	3.2818	25.59	8.034	1.000	9.371	< 0.001
	30	3.7459	3.2908	26.73	7.904	1.000	9.251	< 0.001
	45	3.7543	3.3108	24.99	7.121	1.000	8.456	< 0.001
	60	3.7746	3.2629	22.49	7.45	1.000	8.772	< 0.001
30 kg	0	3.8302	3.2574	22.55	9.999	1.000	11.322	< 0.001
	15	3.8077	3.2422	25.08	8.559	1.000	9.894	< 0.001
	30	3.8143	3.2495	25.08	8.559	1.000	9.894	< 0.001
	45	3.8383	3.2769	25.06	10.184	1.000	8.456	< 0.001
	60	3.8049	3.2667	27.00	8.674	1.000	10.028	< 0.001

Table F.4 The t values and p values for the TOST equivalence tests for frontal plane peak jerk.

Load Condition	Time Point (min)	MoCap Mean	IMU Mean	Degrees of Freedom	Equivalence Test		NHST Test	
					t-value	p-value	t-value	p-value
0 kg	0	3.3598	2.9063	21.14	4.528	1.000	5.845	< 0.001
	15	3.4176	2.8679	22.05	6.583	1.000	7.904	< 0.001
	30	3.4269	2.9113	23.53	5.445	1.000	6.773	< 0.001
15 kg	0	3.4472	2.9287	18.00	5.782	1.000	7.08	< 0.001
	15	3.4763	2.9258	21.02	6.408	1.000	7.724	< 0.001
	30	3.4743	2.9596	17.35	5.709	1.000	7.011	< 0.001
30 kg	0	3.5012	2.9129	23.08	4.751	1.000	6.076	< 0.001
	15	3.4929	2.9707	20.98	4.861	1.000	6.178	< 0.001
	30	3.5191	3.0303	18.56	4.158	1.000	5.465	< 0.001
0 kg	0	3.3598	2.9063	21.14	4.528	1.000	5.845	< 0.001
	15	3.4176	2.8679	22.05	6.583	1.000	7.904	< 0.001
	30	3.4269	2.9113	23.53	5.445	1.000	6.773	< 0.001
15 kg	0	3.4472	2.9287	18.00	5.782	1.000	7.08	< 0.001
	15	3.4763	2.9258	21.02	6.408	1.000	7.724	< 0.001