PROLONGED LOAD CARRIAGE DURING WALKING INDUCES FATIGUE AND REDISTRIBUTES LOWER LIMB MUSCLE EFFORT

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by

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PROLONGED LOAD CARRIAGE DURING WALKING INDUCES FATIGUE AND REDISTRIBUTES LOWER LIMB MUSCLE EFFORT

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LIST OF ABBREVIATIONS

CoM	Center of Mass
EMG	Electromyography
MPF	Mean Power Frequency
rMPF	rate of change of Mean Power Frequency
TA	Tibial Anterior
SOL	Soleus
MG	Medial Gastrocnemius
RF	Rectus Femoris
BF	Biceps Femoris
GMED	Gluteus Medialis
GMAX	Gluteus Maximus

SUMMARY

Load carriage during walking (e.g., a backpack) is prevalent in everyday tasks as well as in industrial and military settings. During loaded walking, the need to support additional weight and maintain dynamic balance induces immediate changes in joint mechanics and lower body muscle activation patterns. The manifestation of acute changes in workload among lower-limb muscles during prolonged, fatiguing walking bouts remains unclear. Based on evidence that load carriage redistributes the mechanical demands toward distal muscles (e.g., ankle plantar flexors) and away from proximal muscles (e.g. hip extensors), we hypothesize that distal muscles would fatigue at a faster rate than proximal muscles. To test this hypothesis, we recruited 8 young healthy adults and compared the rate of change in the mean power frequency (rMPF), a common biomarker of muscle fatigue, for key ankle and hip muscles over a 30 minute walk during both loaded and unloaded conditions. As expected, load carriage caused an immediate increase in recruitment of active muscle volume that was larger for the ankle than for hip muscles. However, contrary to our hypothesis, we found a larger negative rMPF (i.e. more fatigue) for hip extensors (e.g., biceps femoris = -.29 Hz/min) than for the ankle extensors (e.g., soleus = .032 Hz/min) during loaded walking ($p = 0.0022$). This finding suggests the possibility for a motor control strategy that acts to prioritize reliance on proximal muscles during long, highly demanding walking bouts in order to limit fatigue in key distal muscles that are important for efficient propulsive power output.

CHAPTER 1.

1.1 Introduction

During loaded walking, the need to support additional weight and maintain dynamic balance alters joint mechanics and lower body muscle activations patterns. (Sturdy; Mummolo et al.) Vastis lateralis and gastrocnemius muscle activations increased significantly from walking with no load to carrying a load that's equivalent to 20% bodyweight or above (Simpson et al.). Ankle plantar flexor, hip flexors and extensor activities increase significantly during the stance phase and swing phase. (Silder et al.) As a result of altered muscle activation patterns during the stance phase, the increase in peak vertical ground reaction force (GRF) is less than the carried load. (Silder et al.) Peak knee extension angle and peak hip and knee flexion angles increase with increased load during walking. (Loverro et al.) Stride frequency increases as an effort to increase double support time during the gait cycle.(Künzler et al.) Moreover, increased trunk lean makes it more difficult to keep the center of mass (CoM) within the base of support. As a result of both, musculoskeletal stiffness, or segment stiffness induced by muscle, increases during load carriage walking. (Holt et al.) Not only does joint angle increase, but joint moment increases, as well. (Seay et al.) While each joint outputs more mechanical power in comparison to walking with no added load, there seems to be a redistribution of joint power from distal to proximal joints due to the mechanical demand of the activity (Lenton et al.). Unlike a no load walking gait where the ankle joint contributes the most mechanical power (Sawicki and Ferris), loaded walking after ten minutes results in the redistribution of work from the ankle and the knee to the hip (Lenton et al.)

Unequivocally, power redistribution from the ankle to the hip is observed as a training effect, or adaptation strategy, to load carriage activities (Wills et al.), suggesting that load carriage walking has an energy demand profile that's different from no load walking.

While the acute effect of load has been studied, the long term effect of load in an activity session has not been well-characterized. In fact, few studies have touched upon the effect of prolonged locomotion on the neuromuscular system, even without the additional load. During extended walking bouts with a carried load, there is an increased demand for muscle force and work, which can consequently induce neuromuscular fatigue. (Huang and Kuo). Currently, the most prevalent fatigue protocol paradigm includes activities such as repetitions of single joint flexion (Power et al.) or extension (Dalton, Power, Paturel, et al.) or holding for continuous force output at a target percentage of MVC (Kudzia et al.) Less common but also attempted, previous fatigue literature also includes functional activities such as bout of walking or a mixture of walking and other activities, usually in an effort to mimic military tasks. (Kwon et al.) (Grenier et al.) (Zhang et al.)

Muscle activation is routinely reported and interpreted to reflect neuromuscular fatigue, as it's one of the few non-invasive approaches to capture neural activities. In the time domain, compared to baseline, maximum voluntary contraction (MVC) of tibial anterior muscle dropped 28% after repeated high-intensity lengthening contractions (Power et al.). Gluteus maximus and biceps femoris activation increased during the landing phase of a single leg vertical jump.(Lessi et al.) When it comes to measurements taken during walking activities, some reported decreased activation of vastus lateralis, semitendinosus, and medial gastrocnemius (Simpson et al.), while others reported increased activation in ankle and hip muscles as a result of fatigue. (Zhang et al.; Rice et al.) Frequency domain information, such as mean power frequency or median power frequency, has been used to interpret fatigue because fast twitch fibers are more fatigue resistant than slow twitch fibers due to their physiological properties. Muscle activation frequency decreases in plantar flexor muscles (Rice et al.), tibial anterior (Simpson et al.), and all lower limb muscles (Zhang et al.; Kudzia et al.) have been reported and interpreted as indicators of muscular fatigue. Additional characteristics of muscle activation signals have also been analyzed and interpreted for fatigue progression. Fatigue is correlated with a decrease in medium frequency (Zhang et al.), decreased produced force at maximum contraction (O'Leary et al.), a decrease in low frequency torque depression (Power et al.), and a lower rate of muscle activation at medial gastrocnemius (Wallace et al.).

While muscles are actuators whose activations are affected by repetitive use over time, joint kinematics and kinetics are the resulting changes. At the ankle joint, less plantar flexion at heel strike (Oliveira et al.), and decreases in angular velocity, torque at peak power, and peak power (Wallace et al.) indicate that less work is being output at the joint as the fatigue develops. Similar to the ankle joint, peak knee extension moment (Hafer et al.), force produced at maximum voluntary contraction(O'Leary et al.), and knee power (Dalton, Power, Vandervoort, et al.) decreased after the respective fatigue protocol. In contrast to the reduced range of motion of previous joints, increased hip extension and hip flexion have been observed after fatigue. (Hafer et al.) When viewing the entire lower limb as a whole, fatigue impacts leg force control responsiveness but not as significantly on leg force control accuracy (Kudzia et al.), indicating that the mechanical capacity of work output is retained. Similarly, a recent study expanded on the idea of force output control and showed an increased mean mediolateral CoM displacement post-fatigue. (Kwon et al.) When it comes to a more functional task, however, maximum vertical jump height is reduced by 5% before and after fatigue, hinting at an altered lower limb force output pattern before and after fatigue.(O'Leary et al.).

While previous literature has established somewhat consistent metrics for measuring fatigue, most of our knowledge on the topic is from non-locomotive contexts. A full understanding of the redistribution of the joint mechanical power as a result of neuromuscular fatigue remained unclear, as few studies comprehensively addressed interaction between fatigability (i.e.based on electromyography measures) and neuromechanical compensations that result from fatigue. The primary aim of this research is to explore the impact of altered mechanical demand in prolonged human locomotion. More specifically, how increased mechanical demand via load carriage induces neuromuscular fatigue and redistributes work at the joint level. The secondary aim of this research is to add to the locomotion-based fatigue literature by examining how prolonged walking activity induces neuromuscular fatigue at each joint.

We hypothesize that the distal muscles (e.g, ankle plantar flexors) would fatigue at a faster rate than proximal muscles (e.g., hip extensors) over a 31 minute walking bout both with and without carrying load. At minute 0, load carriage induces the greatest increase in ankle power and decrease in hip power at the end of the stance phase. (Huang and Kuo, 2014) Thus, an cumulative effect of more mechanical demand at the ankle joint is expected to fatigue the ankle muscles faster than those around the hip joint. Findings from this research can potentially help to establish reliable metrics for clinicians to monitor fatigue development in rehab patients, for prosthetists to guide the design of active prosthetics that factors the neuromechanics of the intact limb, and for military personnel to develop targeted interventions that are designed to combat fatigue.

1.2 Method

1.2.1 Participants

Eight healthy adults participated in this study after providing informed written consent in accordance with the Georgia Tech Institutional Review Board (average ± SD; six female and two male; age, 21.7 ± 1.2 years; height, 1.65 ± 0.05 m; mass, 65.86 ± 12.18 kg).

1.2.2 Experimental Procedure

Participants walked at their preferred walking speeds on an instrumented split-belt treadmill [Computer-Aided Rehabilitation Environment (CAREN), Motek Medical, The Netherlands] for two sessions (Fig. 1). In one session, the participant walked with no additional load, and in the other with load in the form of a weighted vest. The weight of the vest was rounded to the integer closest to 20% of their body weight. The weight distribution in the vest was balanced in the anterior-posterior and lateral directions.(DATTA and RAMANATHAN) The order of the load carriage conditions is randomized.

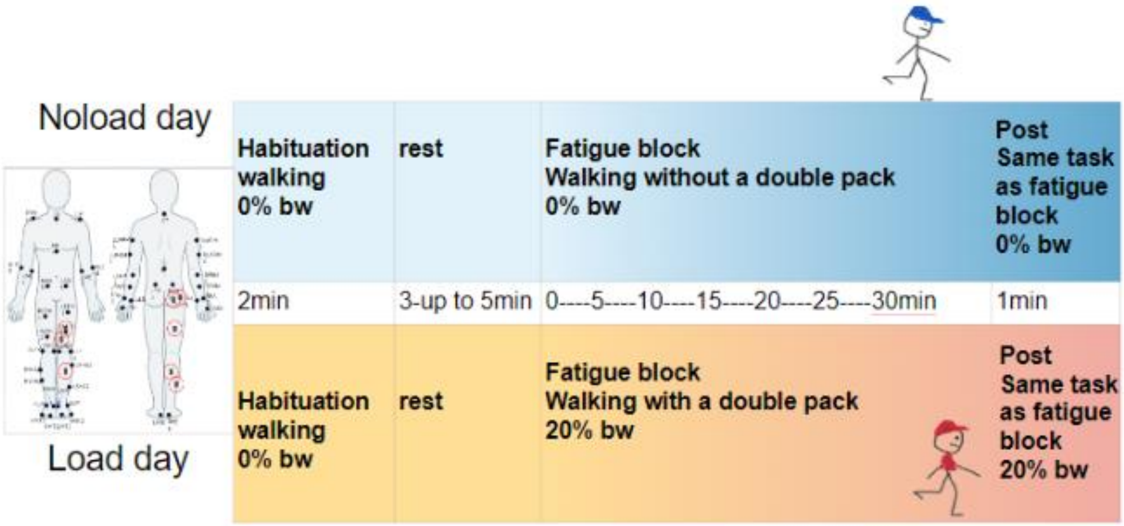

Figure 1.2.2: Experimental Protocol. Left, EMG sensors and reflexive marker placement; right, experiment protocol overview. For load and No Load load days, participants walked on the treadmill for 2 minutes without load where EMG from the second minute was collected as reference for maximum functional MVC, then rested for a minimum of 3 min but no longer than 5 min. For the No Load day, participants continuously walked on the treadmill for 31 minutes without wearing the double pack. For the load day, participants continuously walked on the treadmill for 31 minutes while wearing a weighted vest. Data recorded at the first minute and the 31st minute are compared.

During the first session, their preferred walking speed was obtained by the observer manually adjusting the belt to the participant's preference during the first minute of the acclimation period. The same speed was carried over to the second session. Before getting onto the treadmill, maximum voluntary contraction was collected for a total of eight lower body muscles. Prior to any activity on the treadmill, a static trial and a functional joint center trial is collected in the same condition as the experimental session. I.e. During the load day a double pack is worn in the static trial. During the static trial the participant was asked to remain in a T-pose for 5 seconds and during the functional joint center trial the participant was asked to explore the full range of motion of each lower limb joint. The purpose of the two trials is to facilitate data analysis and the duration of both is less than 60 seconds. During the acclimation period, electromyography (EMG) data was recorded where the second minute was used to determine the maximum functional muscle activation.(Delsys Inc., Natick, MA) Following a two minute treadmill acclimation without wearing a weighted vest, participants continued to walk for another 31 min. While the fatiguing block is 30 minutes in length, the 1-minute-long post fatigue trial took place immediately after the 30-minute-walking without a break to avoid any additional recovery. (Künzler et al.)

1.2.3 Measurements

Muscle activity from eight muscles [tibialis anterior (TA), soleus (SOL), medial gastrocnemius (MG),vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), gluteus medius (GMED) and gluteus maximus (GMAX)] of the dominant leg were recorded using wireless surface electromyography (EMG) (Trigno, Delsys Inc., Natick, MA) at 2000 Hz . EMG sensor is placed on the long head of BF (halfway between its origin at the pelvis and its insertion into the knee joint) to capture its function as an hip extensor. Dominant leg is determined by asking the participant to kick a ball and observing which leg they used. Prior to the placement of each sensor, the exact placement of the sensor is determined by a combination of following the guidance from Neuromechanics of Human Movement and making smaller adjustments to minimize cross talks. (Enoka) In short, we identified the origins and the insertions of each muscle of interest and placed the sensor close to the muscle belly along the fiber length. The skin was cleaned with alcohol wipes and prepared with gel to clear off the dead skins. After the gel was removed with paper towels and the skin was dried, EMG sensors were gently pressed on to ensure good adhesion and secured to the spot with additional medical tapes.

A full body marker set (fig.1 left) was collected by a 10-camera motion capture system (Vicon Motion Systems, UK). During the trials, GRF and EMG data were collected at 2000 Hz while marker data was collected at 100 Hz. For the purposes of this study, we only used the trajectories of the 26 lower-body markers.

1.2.4 Data Processing

1.2.4.1 Joint Mechanics

During the 31 minutes of walking, motion capture, and GRF data was collected every five minutes, starting with minute 1 and ending with minute 31, in two clips of 30 second episodes. Only the first 30 seconds of each five minute period was analyzed because there are usually more than 20 good steps within each 30 second episode of recording. A good step is defined as a step that has a correlation greater than 0.95 to the average shape of the joint angle at all three joints.

Gap-filling is performed using matlab scripts that interfaced with Vicon Nexus. Marker data from the static trial is used to scale each participant's model in OpenSim.(Delp et al.) Each participant has two different models, one each for their session with load or without load, because of the torsal marker (T10 and XPIS) placement difference. Inverse kinematics and inverse dynamics were performed over a total of seven 30s recordings for each session using OpenSim, interfaced with Matlab (Delp et al.) Heel strikes were identified using the kinematic method (Zeni et al.). The processed joint kinematic and kinetic data for each identified stride was then resampled to 3000 points and stride averaged among the good steps. For each participant, joint angle, moment, and power of hip, knee, and ankle joints are characterized by an average stride of all the good steps in the 30s recording.

1.2.4.2 EMG

EMG data was collected during the second minute of the acclimation period to determine the functional maximum activation. During the 31 minutes of walking, EMG data was collected every five minutes, starting with minute one and ending with minute 31, in two clips of 30 second episodes. Only the first 30 seconds of each five minute period was analyzed because there are usually more than 20 good steps within each 30 second episode of recording. In addition to the joint mechanics definition, a good step is further defined by all muscle activations within three times the functional maximum activation of that session.

Data were analyzed in MATLAB (Mathworks, Natick, MA). As the first in post processing, all EMG data are filtered to remove noises from external sources. Similar to other paper that performs frequency domain analysis on muscle, marker trajectories were low pass filtered at 6 Hz and EMG data was bandpass filtered at 15-200Hz (2nd order butterworth), notch filtered at 65Hz, demeaned, rectified, and smoothed using a 200 sample envelope. Secondly, all filtered data for each identified stride was resampled to 3000 points and stride averaged among the good steps. The second minute of the acclimation period EMG data is used to obtain the functional maximum voluntary contraction of each muscle in that session. EMG data for each muscle is stride-averaged over all the good steps in that minute.Then the maximum value in the averaged gait cycle is identified as the functional maximum voluntary contraction for each muscle respectively. Filtered EMG data from 31min continuous walking, with or without load, was normalized using the aforementioned functional maximum voluntary contraction value before further analysis. EMG analysis in the frequency domain was conducted on processed time-series data.

1.2.4.3 Calculation of MPF and rMPF

EMG analysis in the frequency domain was conducted on processed time-series data. Fast Fourier transforms (FFT) were applied to obtain the power spectrum information. The distribution of the frequency over the frequency spectrum was summed and divided by the total number of frequency points to obtain the mean frequency value (MPF). Muscle frequency domain information was interpreted as the primary metric for assessing fatigue because of the size recruitment principle. (Zhang et al.; Kwon et al.)

In order to directly address our hypothesis, mean power frequency rate of change (rMPF) is calculated to enable a direct comparison of the rates of fatigue between muscles. Linear regressions are performed on the set of MPF values for each participant, load or no load conditions, each muscle, from minute 1 to minute 31. That is, rMPF of Gmax muscle for participant 2 during load day is performed by fitting a line with the data consisting of load day minute 1 Gmax MPF, minute 6 Gmax MPF, minute 11 Gmax MPF, …, minute 31 Gmax MFP. The overall slope for each muscle, each condition is calculated by fitting all the participant’s individual data to a line of best fit. For example, the overall rMPF for Gmax muscle during load day is calculated by fitting seven participants’ min 1, min 6, …min 31 Gmax MPF on load day to a line that minimizes the sum of the square of residuals.

Calculations of MPF and rMPF are performed using MATLAB (Mathworks, Natick, MA).

1.2.5 Statistics

In order to test our hypothesis that hip muscles fatigue faster than ankle muscles, we performed two one-way analysis of variance (ANOVA) on the rMPF values. More specifically, the load effect is assessed by determining whether there's a significant difference between the rMPF of minute 1 load and minute 1 no load conditions. And the time effect is assessed by the rMPF min 1 load and min 31 load condition. Comparisons are made for each muscle, respectively. Significance was set at $\alpha=0.05$. Since our primary objective is to assess the time effect of fatigue, further statistical analysis is performed in load condition on hip muscle and ankle muscles. For each participant, the rate of muscular fatigue was obtained by fitting the MPF value of each muscle of interest with a line of best fit. The average line of best fit for each muscle is calculated by fitting multiple pairs of x and y values, namely the MPF values from each participant at every five minute, to the regression. Pair-T tests are performed on factors that yield significance in the ANOVA tests. More specifically, rMPF of BF is compared with rMPF of all the ankle muscles.

To help build rationale for our alternative hypothesis, comparisons between changes in biomechanics metrics were performed. The effects of time and load on several groups of dependent measures were determined using paired-t tests. Load minute 1 data and noload minute 1 data is compared to assess the effect of load. Load minute 1 data and load minute 31 is compared to assess the effect of time, or progression of fatigue. Measures being compared are peak angles, moments, and powers at hip, knee, and ankle joint.

All the statistical processes were performed using custom MATLAB scripts (Mathworks, Natick, MA, USA).

1.3 Result

1.3.1 Joint Kinematics

Load had an immediate effect on joint angles. More specifically, it primarily altered the ankle but not so much the hip joint angles at minute 1. (Fig. 1.3.1.1) Load induced increase in knee flexion during stance and swing phase. Load induced increase in ankle dorsiflexion during the stance phase (0% - 60%). The change in ankle joint angle persisted throughout the 31 minutes, whereas very little change was observed at hip angles, either immediately or over time. (Fig. 1.3.1.2)

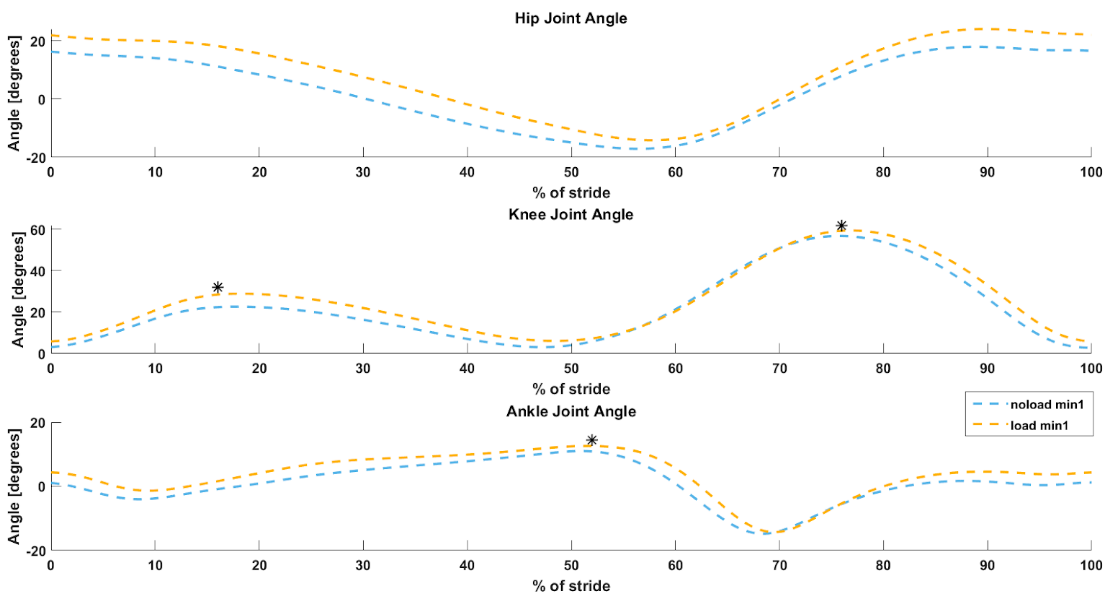

Figure 1.3.1.1: Lower-Limb Joint Kinematics- Load Effect. Ankle (top), knee (middle), hip (bottom) joint angle (degree) over an average stride (0% heel strike to 100% heel strike, same leg) at minute one with (yellow) or without (blue) load. (N = 8) Load induced increase in knee flexion during stance and swing phase. Load induced increase in ankle dorsiflexion during the stance phase (0% - 60%). (N = 8)

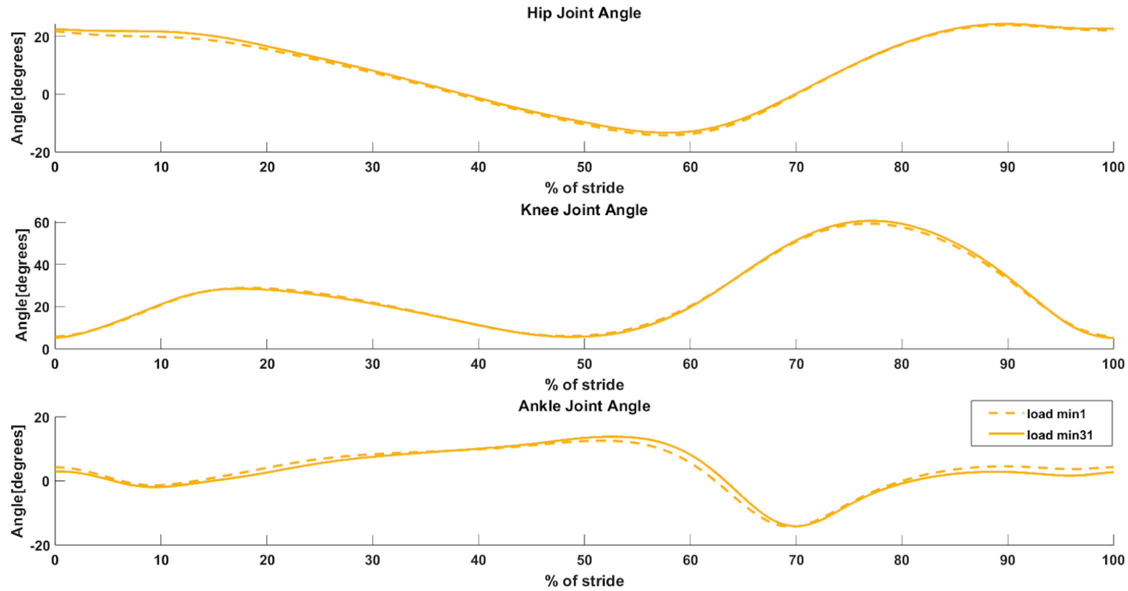

Figure 1.3.1.2: Lower-Limb Joint Kinematics- Time Effect. Ankle (top), knee (middle), hip (bottom) joint angle (degree) over an average stride (0% heel strike to 100% heel strike, same leg) before (dashed line) and after (solid line) the fatigue protocol with (yellow) load. (N = 8)

1.3.2 Joint Kinetics

Load induced an statistically significant increase in hip joint extension moment during the stance phase and increase in hip flexion moment at the push off. Load induced an immediate increase in knee extension moment during the single leg support (10-40%) phase. Load did not induce a significant change in ankle joint moment during the stance phase but seems to have postponed the timing of push off. (Fig. 1.3.2.1) In comparison to the first minute, peak hip extension moment demonstrates a statistically significant increase during the stance phase at minute 31. In comparison to the first minute, the ankle dorsiflexion moment also increased to a less pronounced extent at the push-off at minute 31. Time effect did not alter knee joint moment before and after the fatiguing protocol.

Load induced an instant increase in ankle dorsiflexion flexion, hip flexion, and statistically significant hip extension power. Load induced a statistically significant increase in power absorption at the knee during the single leg support (10-40%) phase. (Fig. 1.3.2.3) While ankle joint power output decreased slightly over the 31 minutes of walking during the stance phase, more positive ankle joint power was observed at the push-off (60%). Oppositely, a decrease in power production and increase in power production was observed at hip during the stance phase and at the push off, respectively. Furthermore, the timing of the push-off was delayed due to load carriage. (Fig. 13.2.4)

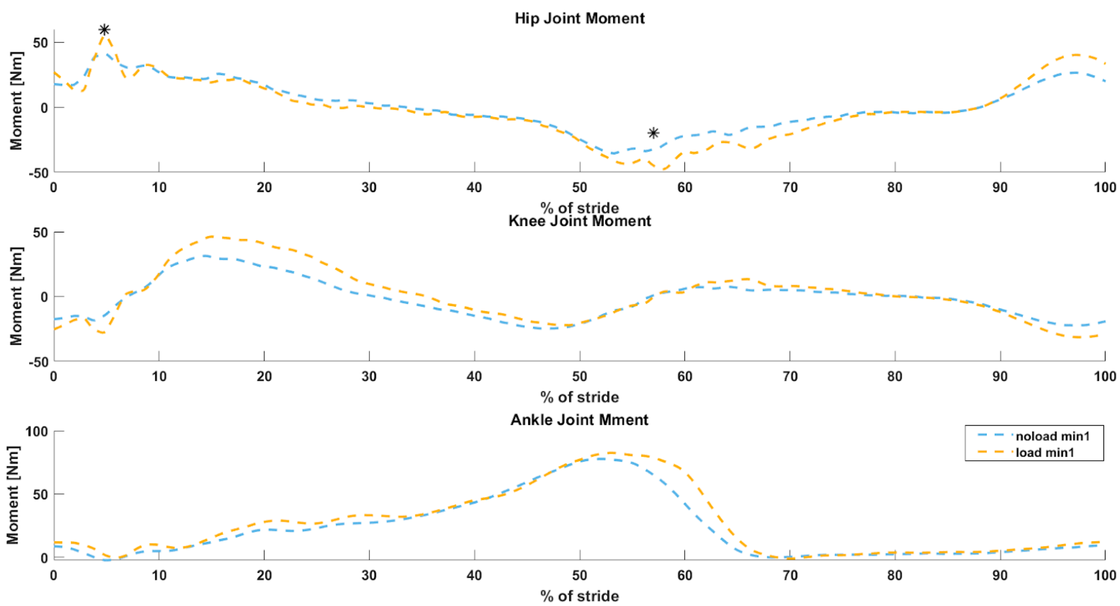

Figure 1.3.2.1: Lower-Limb Joint Kinetics- Moment- Load Effect. Ankle (top), knee (middle), hip (bottom) joint moments (Nm) over an average stride (0% heel strike to 100% heel strike, same leg) at minute one with (yellow) or without (blue) load. (N = 8) Load induced increase in hip extension moment during the stance phase and flexion moment before toe-off.

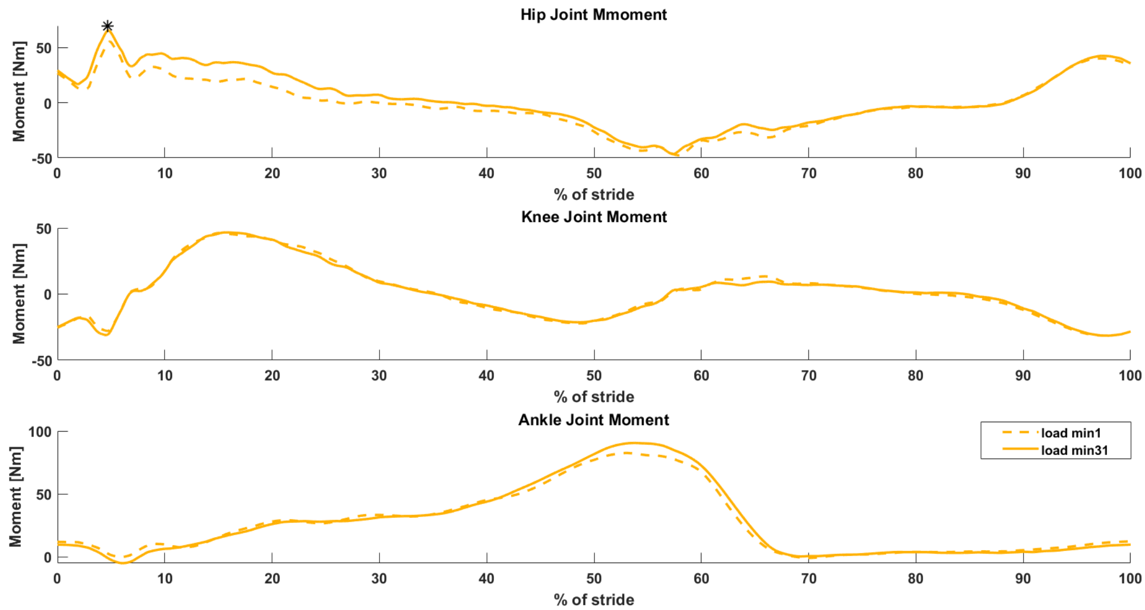

Figure 1.3.2.2: Lower-Limb Joint Kinetics - Moments. Ankle (top), knee (middle), hip (bottom) joint moments (Nm) over an average stride (0% heel strike to 100% heel strike, same leg) before (dashed line) and after (solid line) the fatigue protocol with load. (N = 8) In comparison to the first minute, peak hip extension moment increased during the stance phase at minute 31.

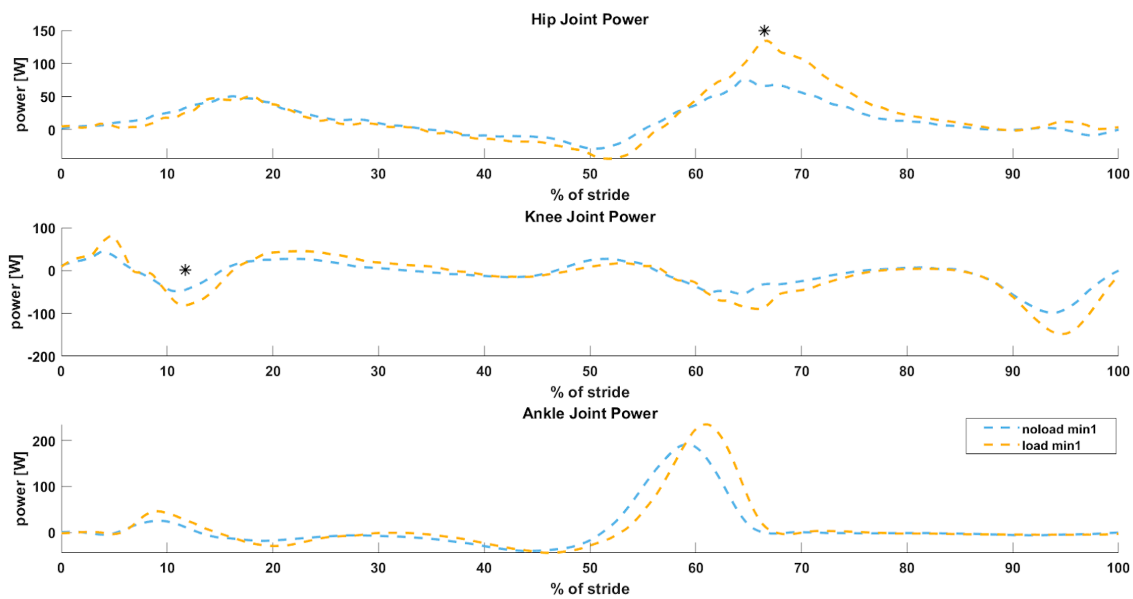

Figure 1.3.2.3: Lower-Limb Joint Kinetics- Powers - Load Effect. Ankle (top), knee (middle), hip (bottom) joint power (N*m) over an average stride (0% heel strike to 100% heel strike, same leg) at minute one with (yellow) or without (blue) load. Load induced increase in hip power generation during the swing phase. (N = 8) Load induced increase in knee power absorption during the stance phase (0% - 60%).

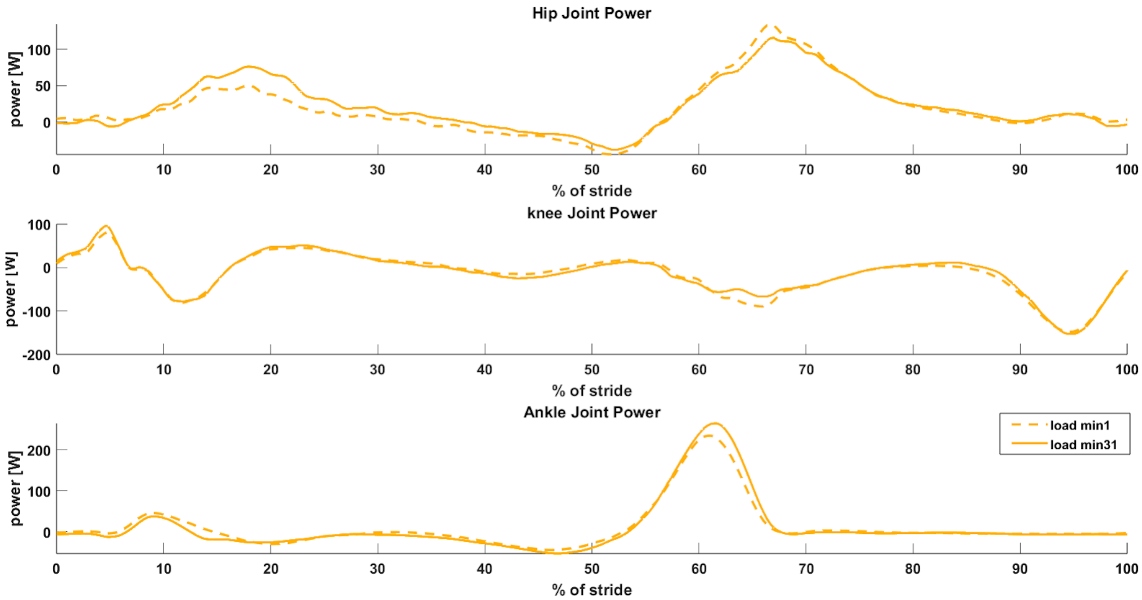

Figure 1.3.2.4: Lower Limb Joint Kinetics - Powers - Time Effect. Ankle (top), knee (middle), hip (bottom) joint power (N*m) over an average stride (0% heel strike to 100% heel strike, same leg) before (dashed line) and after (solid line) the fatigue protocol with load. (N = 8)

1.3.3 Muscle Activation

1.3.3.1 Time Domain

Load had an immediate effect on ankle muscles (ta, soleus) in comparison to all hip muscles. Compared to unload walking, TA and Soleus activation increased by 44.44% and 19.37% at minute 1 due to the addition of load. Ankle muscle activation experienced a minor decrease, whereas hip muscle activation increased to different extents at the end of the loaded walking. No significant changes were observed at either ankle or hip muscles over the duration of unloaded walking.

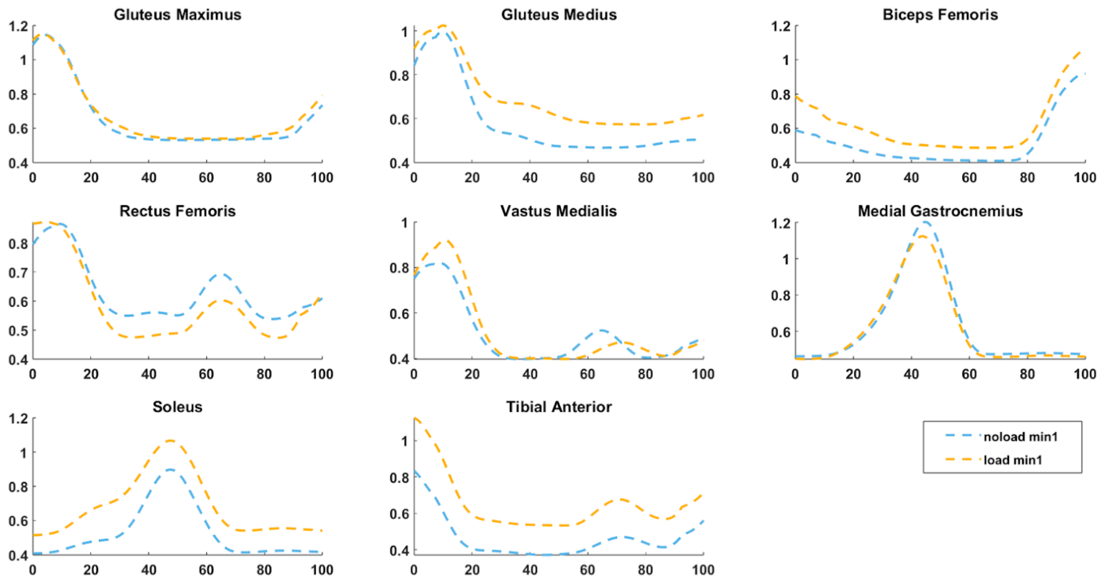

Figure 1.3.3.1.1: Lower-limb Muscle Activity -Time Domain- Load Effect. Magnitude of ankle (Tibial Anterior, Soleus, Medial Gastrocnemius), knee (Vastus Medialis), and hip (Rectus Femoris, Biceps Femoris, Gluteus Medialis, Gluteus Maximus) muscle activations over an average stride (0% heel strike to 100% heel strike, same leg) with (yellow) or without (blue) load, averaged to the maximum functional contraction (N = 7). Load induced significant increase in muscle activation in tibial anterior, soleus, vastus medialis, rectus femoris and biceps femoris.

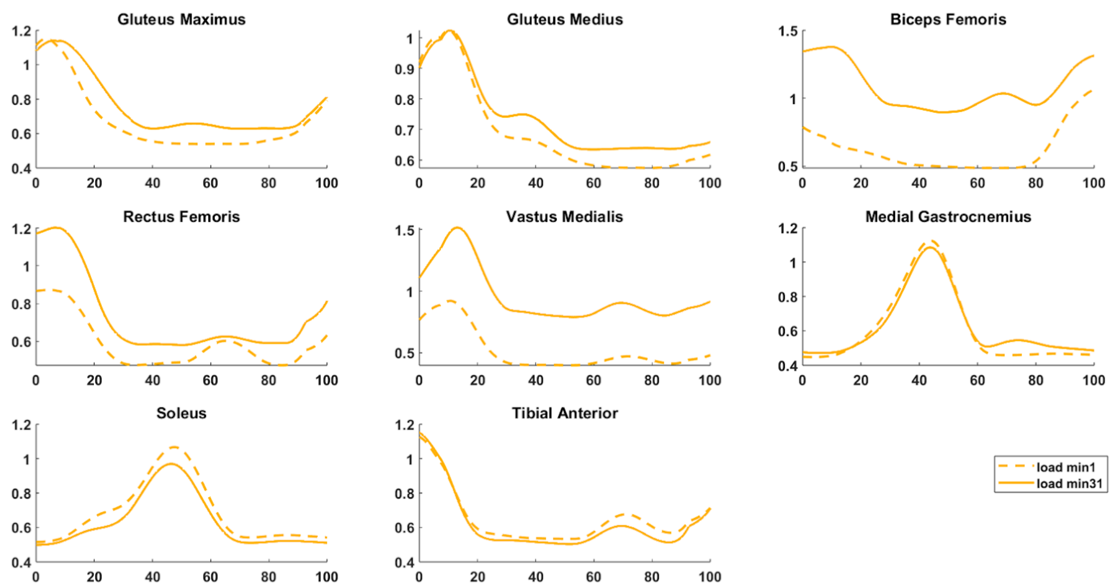

Figure 1.3.3.1.1: Lower-limb Muscle Activity -Time Domain- Time Effect. Magnitude of ankle (Tibial Anterior, Soleus, Medial Gastrocnemius), knee (Vastus Medialis), and hip (Rectus Femoris, Biceps Femoris, Gluteus Medialis, Gluteus Maximus) muscle

activations over an average stride (0% heel strike to 100% heel strike, same leg) before (dashed line) and after (solid line) the fatigue protocol with load, averaged to the maximum functional contraction. (N = 7.)

1.3.3.2 Frequency Domain

Load induced different patterns of change in rMPF at different muscles. rMPF bicep femoris (-0.293 Hz/min) was significantly smaller than all the ankle muscles, namely tibial anterior (0.069 Hz/min), soleus (0.032 Hz/min), and medial gastrocnemius (0.043 Hz/min), during load day ($p = 0.0022$). (Fig. 1.3.3.2.3) No statistical significance was found between hip and ankle muscles during the unloaded day. The individual linear regression results for each participant, each muscle, at each condition is recorded in Appendix A.

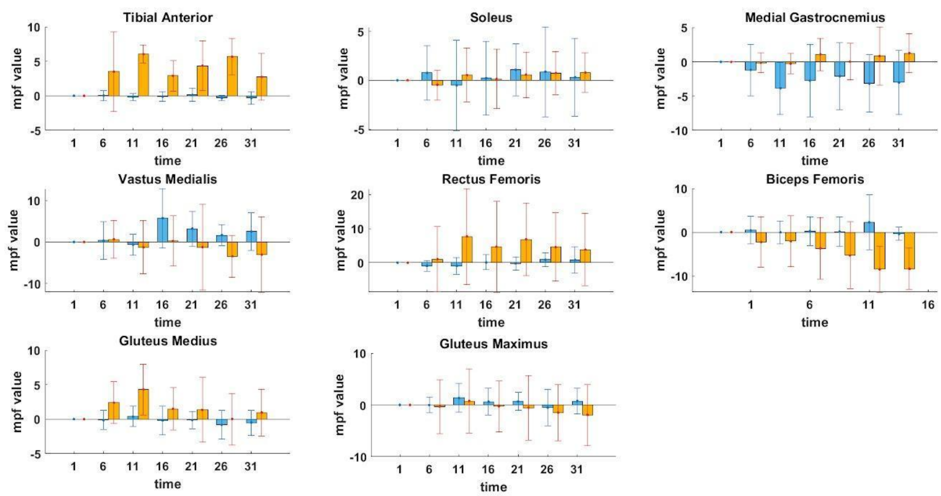

Figure 1.3.3.2.1: Lower-Limb Muscle Activity - Frequency Domain. Changes in mean power frequency (MPF, in Hz) in ankle (tibial anterior, soleus, medial gastrocnemius), knee (vastus medialis), hip (rectus femoris, biceps femoris, gluteus medialis, gluteus maximus) muscles over the duration of 31 minutes of walking with (yellow) or without (blue) load. (N = 7) BF MPF decreased over the duration of 31 min walking with load. Increase in TA rMPF sustained over the duration of 31min walking with load. And decrease in medial gastroc sustained over the duration of 31 min walking without load.

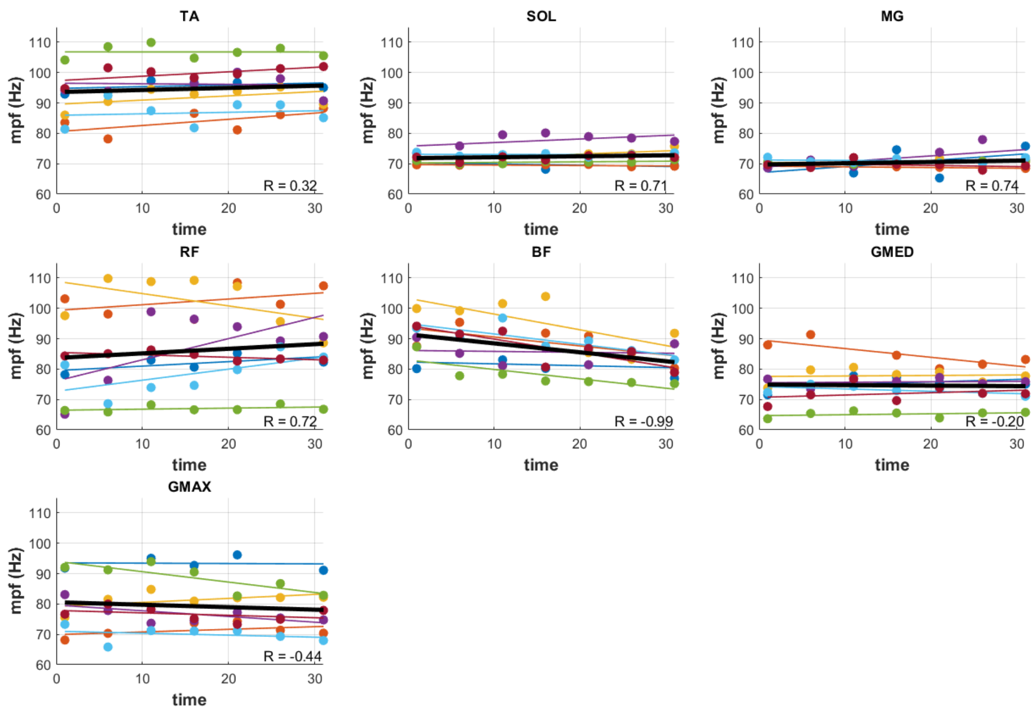

Figure 1.3.3.2.2: Lower-Limb Muscle Activity - Frequency Domain over time with line of best fit. mean power frequency (MPF, in Hz) in ankle (tibial anterior, soleus, medial gastrocnemius) and hip (rectus femoris, biceps femoris, gluteus medialis, gluteus maximus) muscles over the duration of 31 minutes of walking with load.(N = 7) Each color represents one participant and their rMPF, derived by performing linear regression on the scatter plot data points. The Bolden black line represents the mean rMPF across participants for each muscle. The displayed R value shows the strength of correlation between time and MPF value. rMPF of BF has a significantly different slope than all ankle muscles

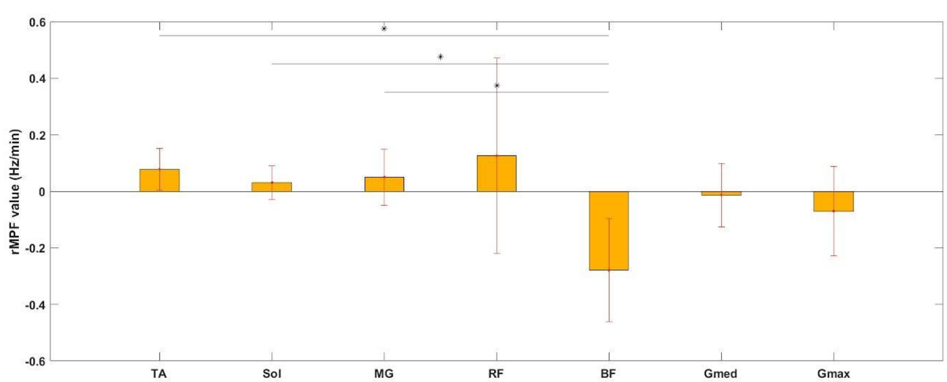

Figure 1.3.3.2.3: Lower-Limb Muscle Activity - Frequency Domain Rate of Change. Mean power frequency rate of change (rMPF, in Hz/min) over 31min during load day. Each bar represents the rate of change of one lower limb muscle. (N = 7)Ankle muscles

are tibial anterior, soleus, medial gastrocnemius. Hip muscles are rectus femoris, biceps femoris, gluteus medius, and gluteus maximus. rMPF of biceps femoris is significantly more negative than all the ankle muscles.

1.4 Discussion

In spite of the prevalence of load carriage in the activity of daily living, very little is known about how load carriage induces and redistributes fatigue in lower limb muscles. For example, while the hip and ankle are known to be the major contributors during walking, it's unclear how the acute changes in workload of lower-limb muscles manifests during extended fatiguing walking. The purpose of this study was to make a direct comparison between muscles at the hip joint and the ankle joint. Just as in previous literature on fatigue, we used MPF as our major metric, and a decrease in MPF is interpreted as the result of muscular fatigue. (Kwon et al.; Zhang et al.) Because mechanical demand at the distal muscle is greater than that at the proximal muscle when carrying load (Silder et al.), we expected that load carriage would disproportionately increase mechanical demand on the ankle, leading to fatigue developing faster in distal muscles (ankle muscles) than in proximal muscles (hip muscles).

Contrary to the proposed hypothesis, our data indicate that proximal muscles fatigue faster than distal muscles during load carriage walking. Namely, MPF of biceps femoris (BF), a hip extensor muscle, decreased significantly faster than medial gastroc, soleus, and tibial anterior over the 31 minute walk while carrying load (Fig. 6). While we are aware that BF is a bi-articular muscle and functions as a knee flexor and a hip extensor in locomotion, the placement of EMG electrode between two attachment points of the long head specified it to be a measurement of its hip extensor function. Among ankle muscles, no significant decrease in MPF was observed. While there is a consistent decrease in

MPF of medial gastroc activation during no load walking session, no pattern of decrease over time was observed at the ankle muscles. (fig 6 - gastro) These findings were unexpected, given that an acute increase in ankle dorsiflexion (Fig. 2a,) during the late stance phase due to load was observed. Despite no significant changes in peak hip joint angle, load induced significant increase in peak flexion and extension moment at the hip as well as power generation during at the onset of single leg support. (Fig 3a, Fig 4a) The observation increase in joint extension moment and positive power at the hip joint partially aligned with previous research regarding the immediate effect of different magnitude of load on gaits. (Lenton et al. Silder. et al) More specifically, the proportion of work performed by the hip joint increased as the magnitude of load carriage increased. At the ankle joint, however, despite a significant increase in the ankle dorsiflexion angle which aligned with previous load carriage studies, no statistical difference is observed in ankle moment changes. (Huang et al.; Silder et al.) Evidence from the time series muscle activation data also substantiated an increase in mechanical demand due to load. Similar to a previous study, an increase in tibial anterior and soleus muscle efforts was elicited (Fig. 5) by load. (Paul et al.)

It is unclear why the loaded walking did not fatigue ankle muscles despite increasing the mechanical demand at the onset. At the motor neuron level, there might be a motor control strategy that limits the total work done by the distal muscles in order to preserve their ability to contract. This is evident in the observed decrease in all ankle muscle activation after the 30 minute loaded walk (Fig. 5). Instead of recruiting fast twitch fibers to keep up with the initial activation level, which should be reflected by an increase in

MPF, there was no significant shift in MPF at soleus and medial gastroc, the muscles that perform during the push-off, from their respective minute 1 values during load day. (Fig. 6) This indicated that no additional fast twitch fibers were recruited to complement the force output capacity of distal muscles. These observations point to a control strategy that inhibits the exertion of distal muscle during load carriage despite a decrease in the performance of the original slow twitch fiber. Since ankle muscles play a vital role in arresting from fall, this neural control strategy to contain muscular effort at the ankle joint may be related to how different types of motor tasks are prioritized. Ankle muscles are activated when the body's CoM deviates from the center of pressure to restore balance. (Reimann et al.)

Another possible reason for the increased MPF in TA (ankle dorsiflexor) and RF (hip flexor) starting at minute 6 is the function of flexors and extensors in regulating stance time. Load carriage increased stance time, as mentioned in previous literature (Park et al.; Künzler et al.)(Fig 2a and Fig 4a), and flexors of each joint are involved in delaying the toe-off in the context of load carriage walking. Additionally, TA peak fascicle lengthening velocity is involved in regulating walking to achieve a more metabolically efficient speed.(Kwak and Chang) Hence, the elevated activation in TA might reflect an effort in walking-speed regulation. Future studies could investigate the effect of walking speed on fatiguing effects with an emphasis on comparing flexors and extensors at the muscle physiology level (i.e. looking at the fiber length/contraction velocity change).

One of the limitations in our study is the measurement of the preferred walking speed. In our protocol, the preferred walking speed was assessed during the acclimation period of the first session, and no load was involved in the acclimation period for both loaded and unloaded sessions. The preferred walking speed during loaded walking might be slower because of the need for a longer stance phase. Walking at a faster speed might elicit a muscle level demand to regulate for faster pacing. Additionally, no qualitative feedback such as Borg rating scale of perceived exertion or muscle soreness rating was collected. Although the absence of qualitative data does not interfere with the significance of this study, they would have been a good real world reference to the relative muscle activation we observed in the study. Furthermore, participants are of different fitness levels, so a universal 20% bodyweight double pack load might induce more fatigue in those who are less physically adapted than those who are more physically fit. While participants from our study are a reasonable representation of the healthy young adult population, future studies can aim to tailor fatigue protocols that are based on an individual's fitness level or even focusing on a subgroup of population with similar fitness levels to increase the significance of their results.

1.5 Conclusion

Our study is the first to investigate the effect of load on fatigue redistribution between muscles. In contradiction to our hypothesis,, we found a larger negative rMPF (i.e. more fatigue) for hip extensors (e.g., biceps femoris = -.29) than for the ankle extensors (e.g., soleus = .032) during loaded walking ($p = 0.0022$). This finding suggests the possibility of a motor control strategy that acts to prioritize reliance on proximal muscles during long, highly demanding walking bouts in order to limit fatigue in key distal muscles that are important for efficient propulsive power output and balance regulation.

APPENDIX A. RMPF FOR EACH PARTICIPANT EACH MUSCLE EACH CONDITION

Table A. rMPF for each participant, each muscle, each condition.

	Load		No load	
	individual	overall slope (R value)	individual	overall slope/R
TA	0.0582		-0.0389	
	0.0169		0.000762	
	0.138		0.00432	
	0.0182		-0.0264	
	0.000487		-0.0245	
	0.0500		0.0133	
	0.146	0.069 (.32)	-0.0162	-0.022 (.13)
SOL	-0.0408		-0.0489	
	-0.0159		-0.0302	
	-0.103		-0.0633	
	-0.117		0.0794	
	-0.0191		0.124	
	-0.0027		-0.113	
	0.0353	0.032 (.71)	-0.181	-0.0081 (-.056)
MG	0.200		-0.0692	
	-0.0192		-0.00747	
	0.0167		-0.327	
	0.188		-0.0533	
	0.0164		0.0829	
	-0.0127		--0.151	
	-0.0374	0.043 (.74)	-0.0406	-0.11 (-.76)

Table A continued

	Load		No load	
	individual	overall slope (R value)	individual	overall slope/R
RF	0.147		-0.0806	
	0.122		0.236	
	-0.406		-0.00452	
	0.698		-0.00977	
	0.0342		-0.0503	
	0.364		0.0416	
	-0.0802	0.15 (.72)	0.0888	0.06 (.55)
BF	-0.0609		-0.0686	
	-0.245		0.201	
	-0.518		-0.00543	
	-0.0304		-0.0186	
	-0.309		-0.0890	
	-0.340		0.00772	
	-0.449	-0.29 (-.99)	0.0717	0.027 (0.23)
GMED	0.0834		-0.0325	
	-0.239		-0.130	
	0.0166		-0.00588	
	0.0158		0.0142	
	0.0308		-0.0496	
	-0.0769		0.0173	
	0.0766	-0.017 (-.20)	0.0440	-0.019 (-0.78)
GMAX	-0.0107		-0.0211	
	0.0619		-0.126	
	0.131		0.0137	
	-0.188		0.0907	
	-0.342		-0.0442	
	-0.0649		0.0714	
	-0.0799	-0.08 (-.44)	0.0535	0.0016 (0.38)

APPENDIX B. EXEMPLARY EMG DATA PROCESSING: S4 SOLEUS NO LOAD CONDITION MINUTE 1

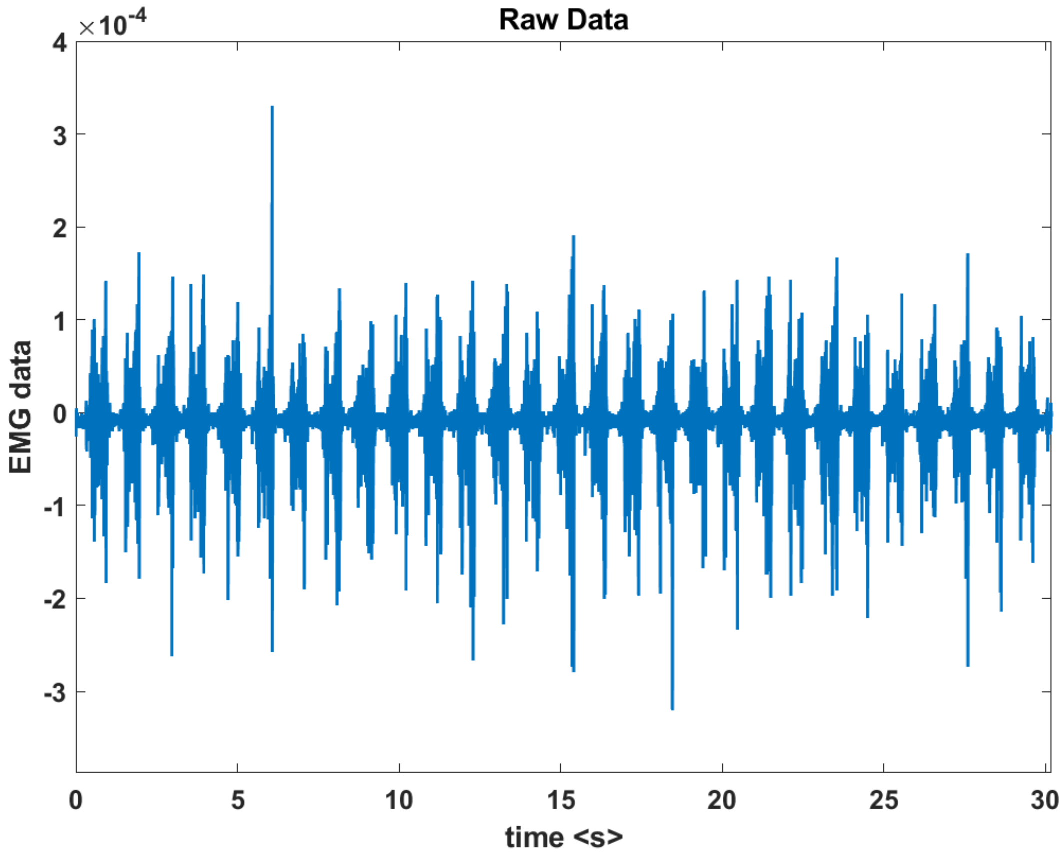

Fig. B.1: Example of Raw Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4.

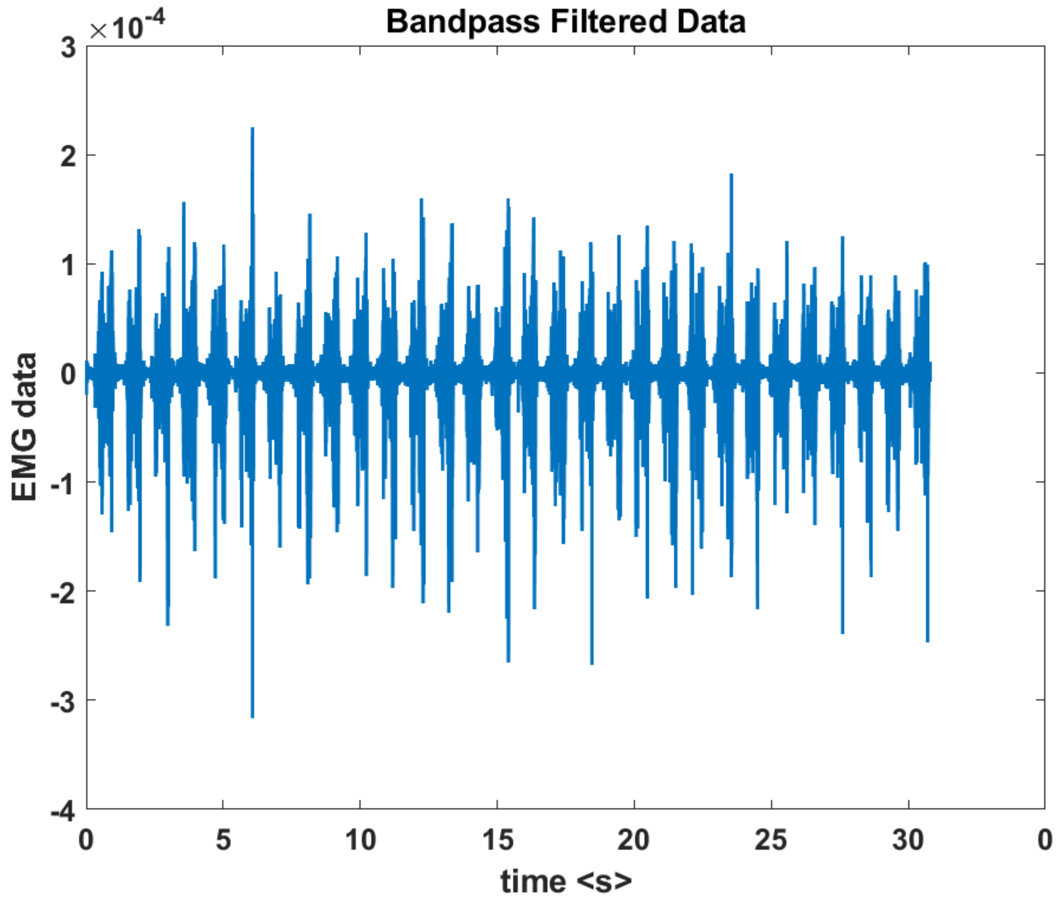

Fig. B.2: Example of Bandpass Filtered Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The raw data was bandpass filtered with a second order butterworth filter with frequency cutoff at 15Hz and 200Hz.

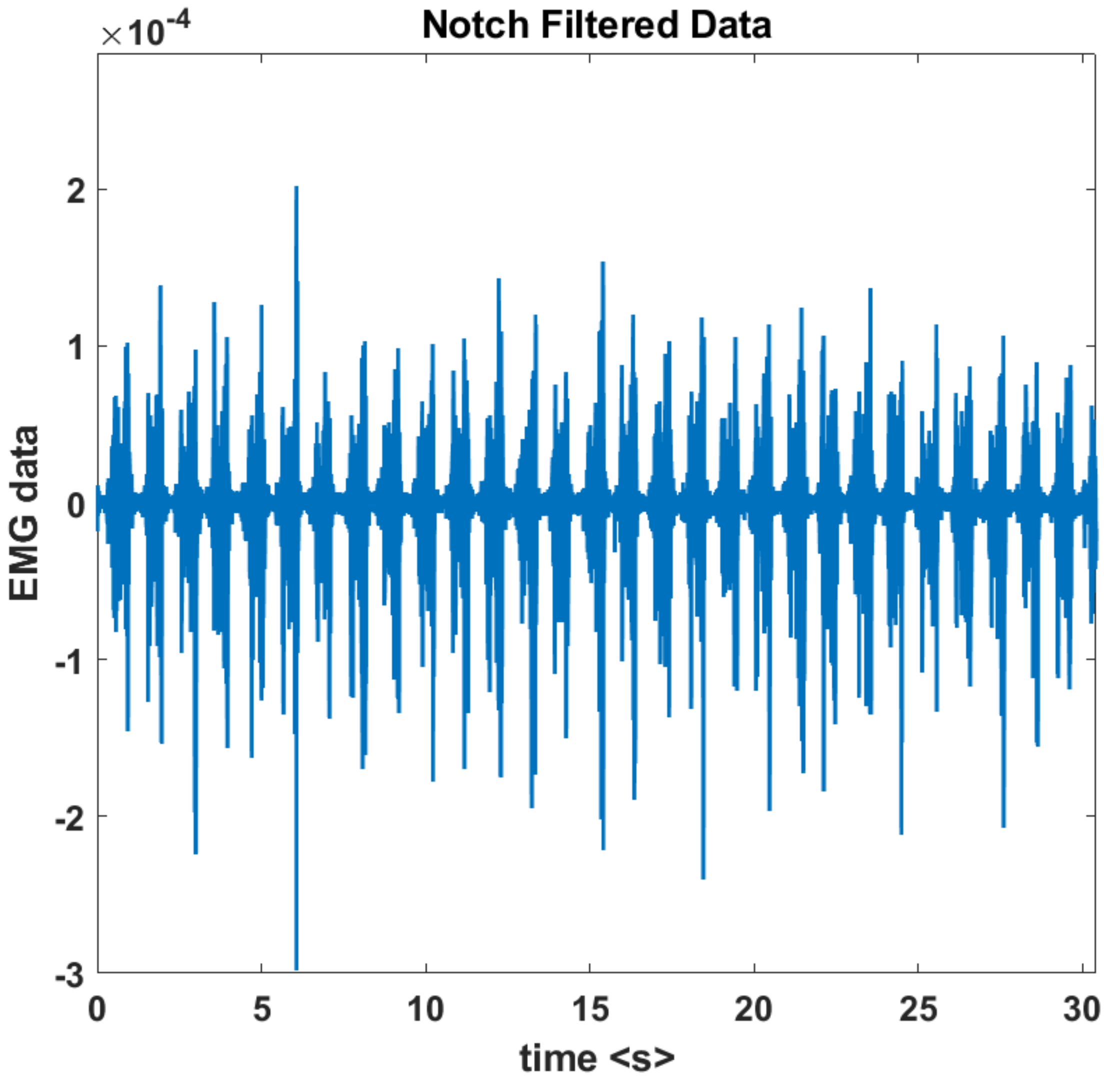

Fig. B.3: Example of Notch Filtered Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The bandpass filtered data was notch filtered to remove AC artifacts between 55 - 65 Hz.

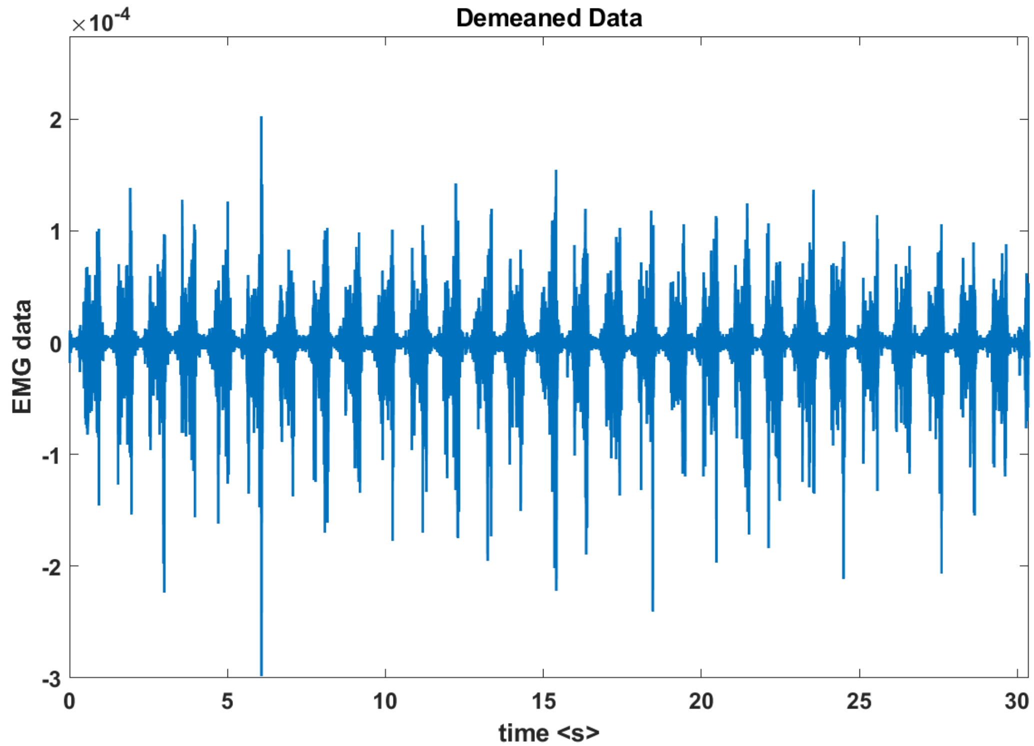

Fig. B.4: Example of Demeaned Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The filtered data was subtracted by its mean value to be centered around zero.

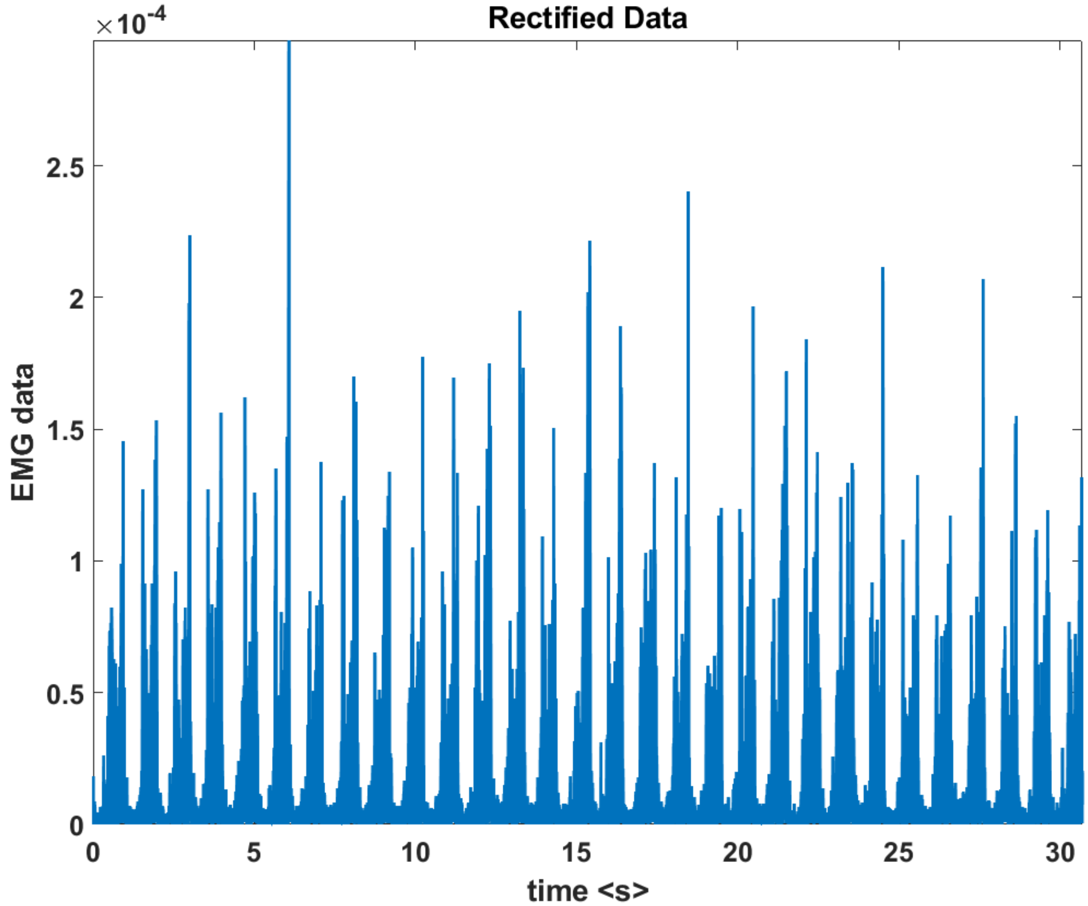

Fig. B.5: Example of Rectified Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The demeaned data was rectified by taking the absolute value of all data.

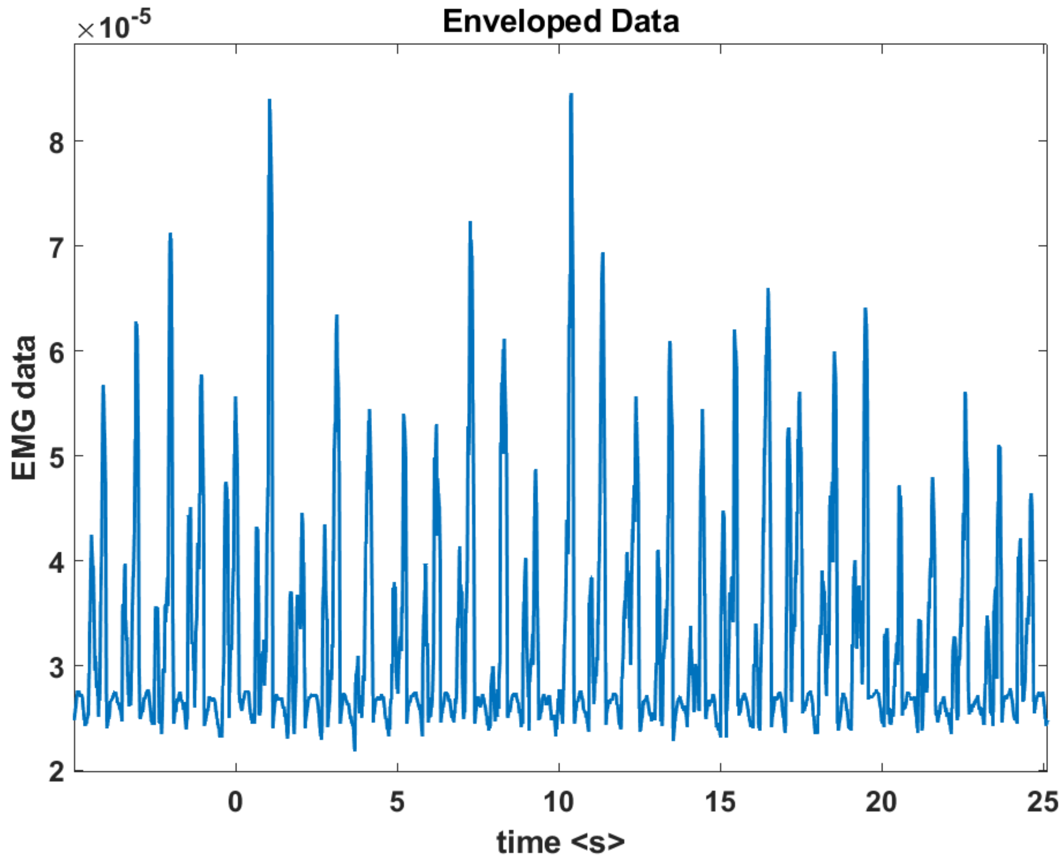

Fig.B.6: Example of Enveloped Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The rectified data was smoothed using RMS method over a sliding window of 200 samples.

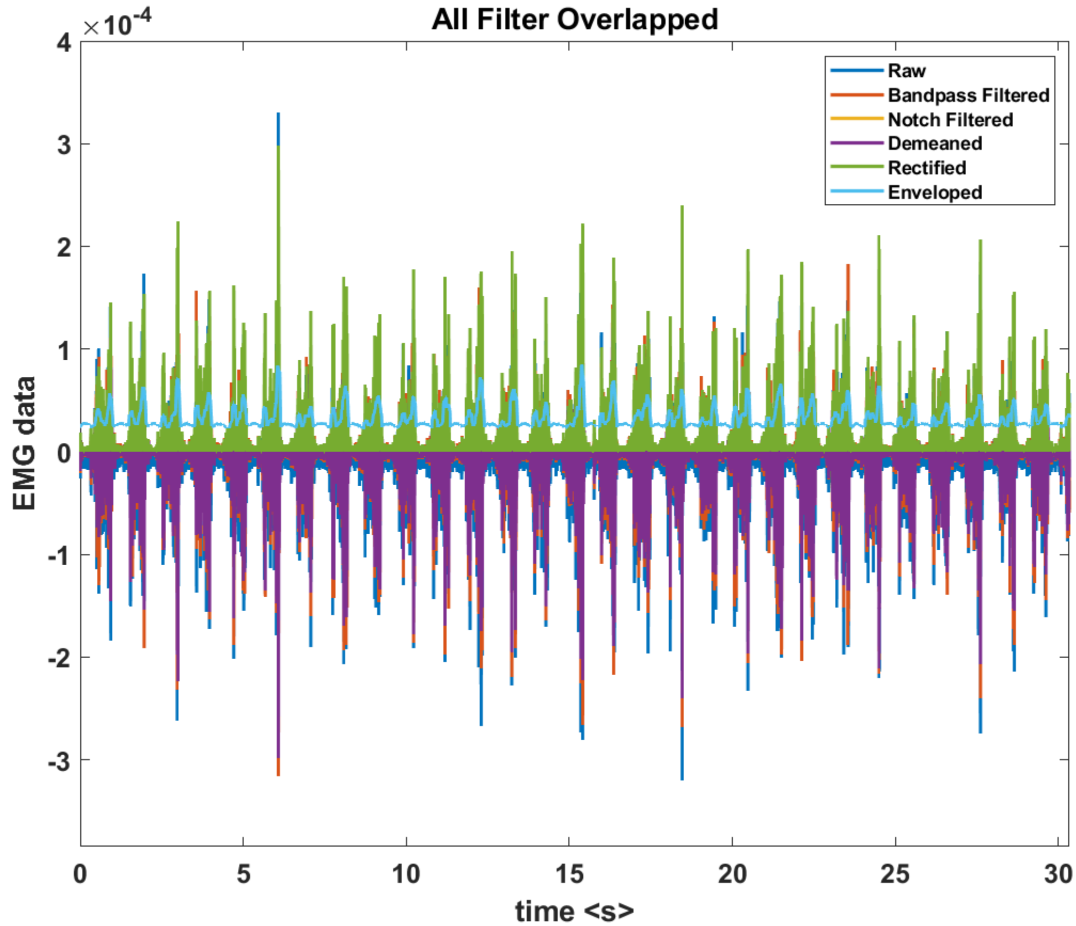

Fig. B.7: Example of Time Serie Muscle Activation Data Processing. First minute no load condition soleus muscle activation from participant #4. The step to step processing was overlaid with legend.

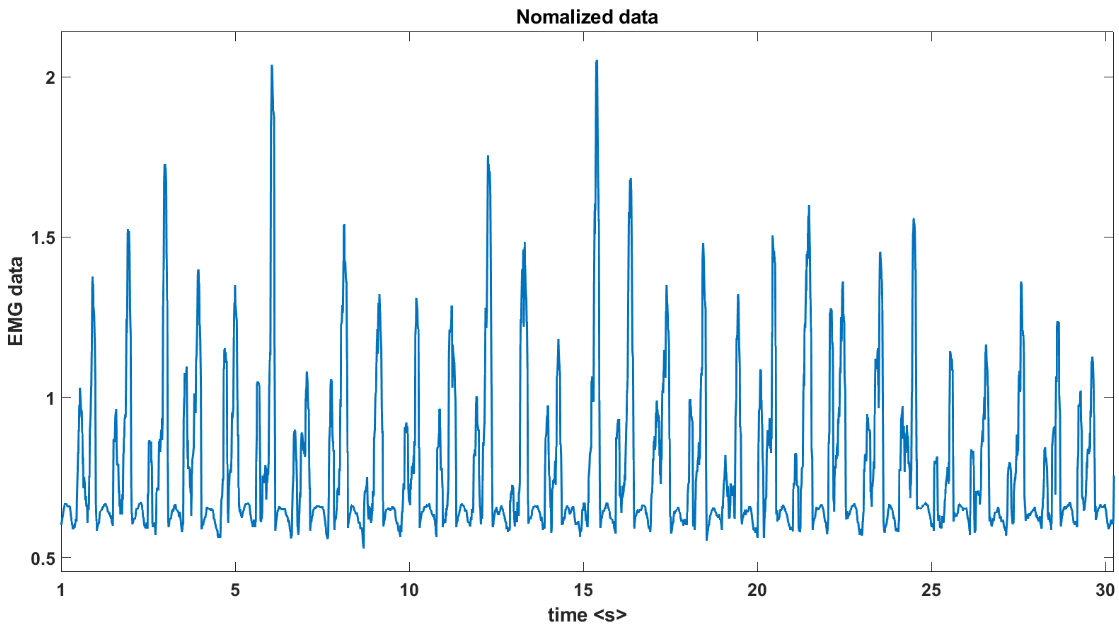

Fig. B.8: Example of Normalized Time Serie Muscle Activation Data. First minute no load condition soleus muscle activation from participant #4. The processed data was normalized to the maximum activation in an averaged gait cycle activation during the second minute (in the habituation phase)

APPENDIX C. EXEMPLARY MECHANICS DATA GRAPH: S4

SOLEUS LOAD CONDITION MINUTE 1

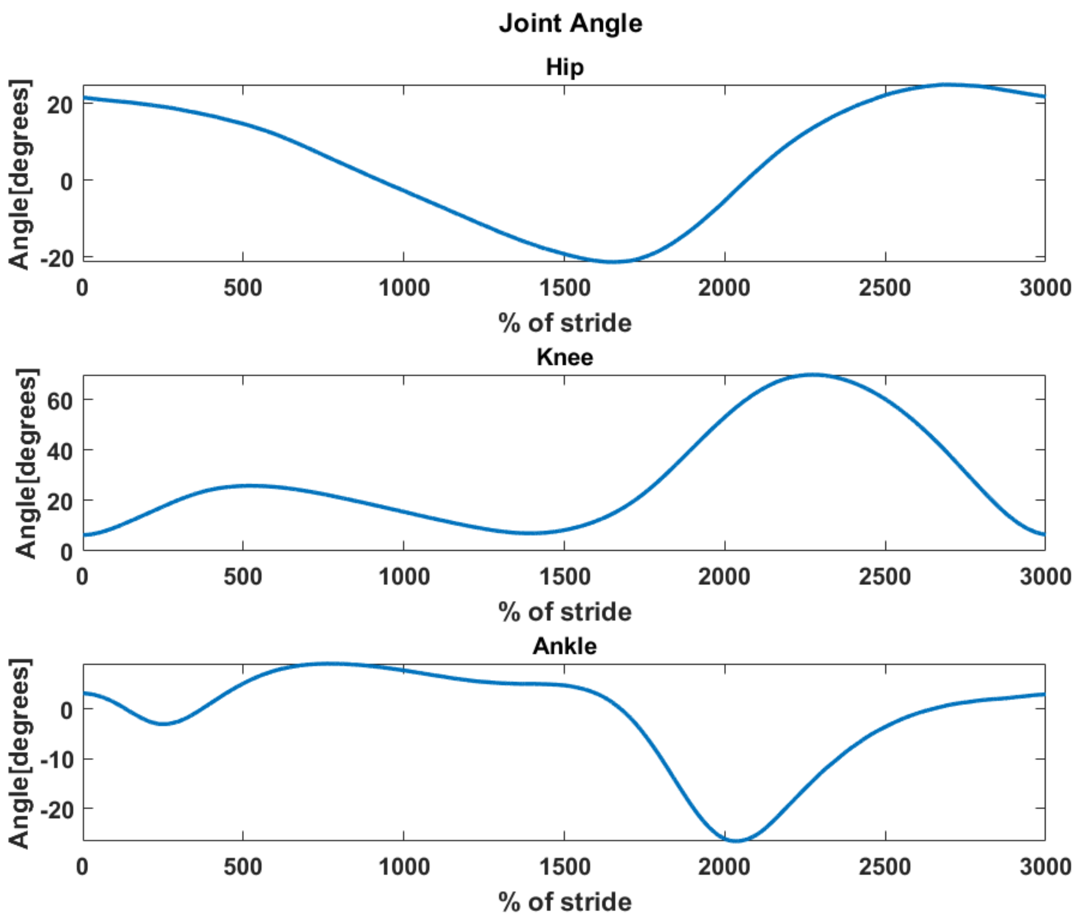

Fig. C.1: Example of Joint Kinematics Data. First minute load condition hip, knee, ankle joint angle averaged over an average stride (0% heel strike to 100% heel strike, same leg) from participant #4.

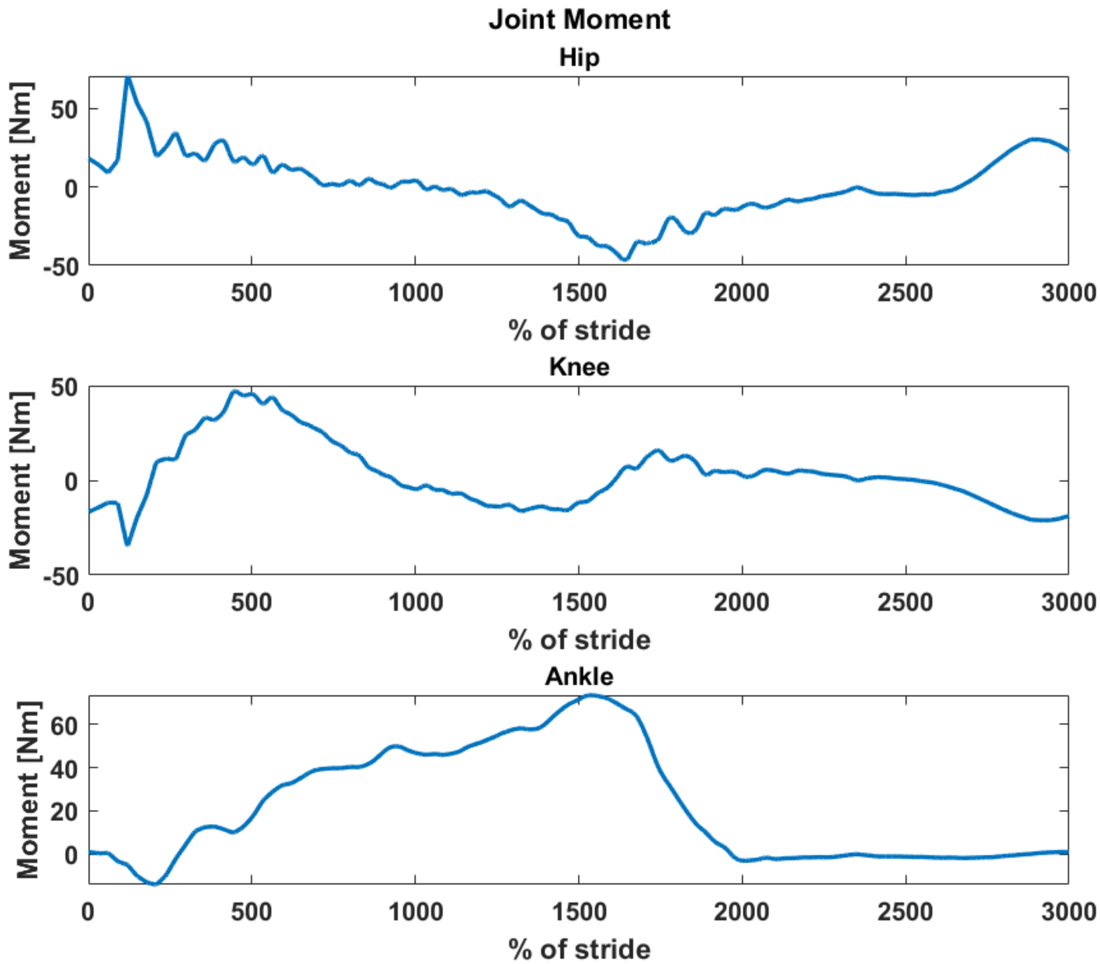

Fig. C.2: Example of Joint Kinetics Data. First minute load condition hip, knee, ankle joint moment averaged over an average stride (0% heel strike to 100% heel strike, same leg) from participant #4.

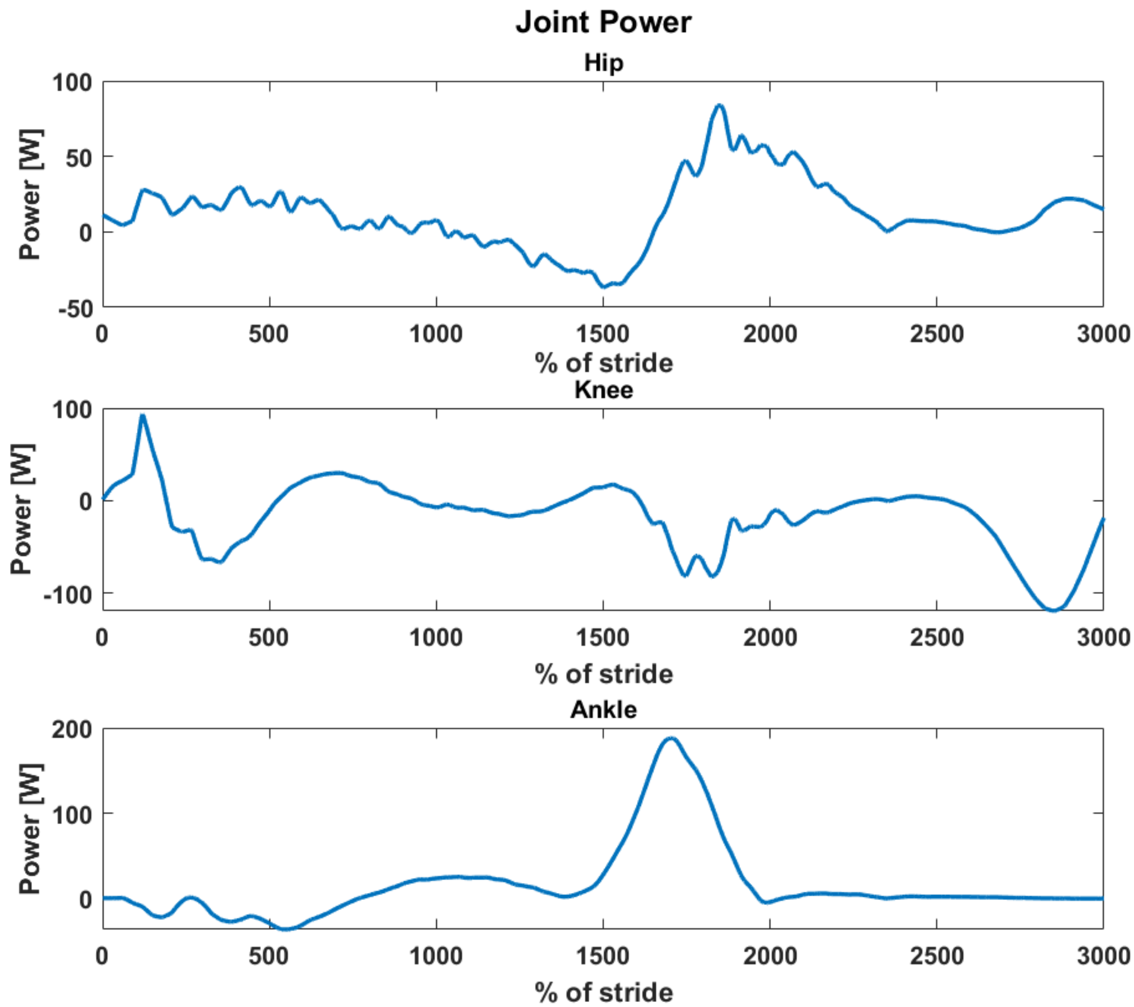

Fig. C.3: Example of Joint Kinetics Data. First minute load condition hip, knee, ankle joint power averaged over an average stride (0% heel strike to 100% heel strike, same leg) from participant #4.

REFERENCES

Dalton, Brian H., Geoffrey A. Power, Justin R. Paturel, et al. “Older Men Are More Fatigable than Young When Matched for Maximal Power and Knee Extension Angular Velocity Is Unconstrained.” Age (Dordrecht, Netherlands), vol. 37, no. 3, June 2015, p. 9790. PubMed, https://doi.org/10.1007/s11357-015-9790-0.

Dalton, Brian H., Geoffrey A. Power, Anthony A. Vandervoort, et al. “The Age-Related Slowing of Voluntary Shortening Velocity Exacerbates Power Loss during Repeated Fast Knee Extensions.” Experimental Gerontology, vol. 47, no. 1, Jan. 2012, pp. 85–92. PubMed, https://doi.org/10.1016/j.exger.2011.10.010.

DATTA, S. R., and N. L. RAMANATHAN. “Ergonomic Comparison of Seven Modes of Carrying Loads on the Horizontal Plane.” Ergonomics, vol. 14, no. 2, Mar. 1971, pp. 269–78. Taylor and Francis+NEJM, https://doi.org/10.1080/00140137108931244.

Delp, Scott L., et al. “OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement.” IEEE Transactions on Biomedical Engineering, vol. 54, no. 11, Nov. 2007, pp. 1940–50. IEEE Xplore, https://doi.org/10.1109/TBME.2007.901024.

Enoka, Roger M. Neuromechanics of Human Movement. Human Kinetics, 2008.

Grenier, Jordane G., et al. “Effects of Extreme-Duration Heavy Load Carriage on Neuromuscular Function and Locomotion: A Military-Based Study.” PLOS ONE, vol. 7, no. 8, Aug. 2012, p. e43586. PLoS Journals, https://doi.org/10.1371/journal.pone.0043586.

Hafer, Jocelyn F., et al. “The Roles of Sex and Physical Activity in Gait and Knee Extensor Function With Age.” Journal of Applied Biomechanics, vol. 35, no. 4, Aug. 2019, pp. 263–71. journals.humankinetics.com, https://doi.org/10.1123/jab.2018-0406.

Holt, Kenneth G., et al. “Increased Musculoskeletal Stiffness during Load Carriage at Increasing Walking Speeds Maintains Constant Vertical Excursion of the Body Center of Mass.” Journal of Biomechanics, vol. 36, no. 4, Apr. 2003, pp. 465–71. ScienceDirect, https://doi.org/10.1016/S0021-9290(02)00457-8.

Huang, Tzu-wei P., and Arthur D. Kuo. “Mechanics and Energetics of Load Carriage during Human Walking.” Journal of Experimental Biology, vol. 217, no. 4, Feb. 2014, pp. 605–13. Silverchair, https://doi.org/10.1242/jeb.091587.

Kudzia, Pawel, et al. Neuromuscular Fatigue Reduces Responsiveness When Controlling Leg External Forces. bioRxiv, 25 May 2023, p. 2023.05.24.541485. bioRxiv, https://doi.org/10.1101/2023.05.24.541485.

Künzler, Marina, et al. “Effect of Load Carriage on Joint Kinematics, Vertical Ground Reaction Force and Muscle Activity: Treadmill versus Overground Walking.” Gait & Posture, vol. 104, July 2023, pp. 1–8. ScienceDirect, https://doi.org/10.1016/j.gaitpost.2023.05.018.

Kwak, Samuel T., and Young-Hui Chang. “Fascicle Dynamics of the Tibialis Anterior Muscle Reflect Whole-Body Walking Economy.” Scientific Reports, vol. 13, no. 1, 1, Mar. 2023, p. 4660. www.nature.com, https://doi.org/10.1038/s41598-023-31501-2.

Kwon, Yujin, et al. “The Effect of Prolonged Walking on Leg Muscle Activity Patterns and Vulnerability to Perturbations.” Journal of Electromyography and Kinesiology, Nov. 2023, p. 102836. ScienceDirect, https://doi.org/10.1016/j.jelekin.2023.102836.

Lenton, Gavin K., et al. “Lower-Limb Joint Work and Power Are Modulated during Load Carriage Based on Load Configuration and Walking Speed.” Journal of Biomechanics, vol. 83, Jan. 2019, pp. 174–80. ScienceDirect, https://doi.org/10.1016/j.jbiomech.2018.11.036.

Lessi, Giovanna Camparis, et al. “Effects of Fatigue on Lower Limb, Pelvis and Trunk Kinematics and Muscle Activation: Gender Differences.” Journal of Electromyography and Kinesiology, vol. 32, Feb. 2017, pp. 9–14. ScienceDirect, https://doi.org/10.1016/j.jelekin.2016.11.001.

Loverro, Kari L., et al. “Females and Males Use Different Hip and Knee Mechanics in Response to Symmetric Military-Relevant Loads.” Journal of Biomechanics, vol.

95, Oct. 2019, p. 109280. DOI.org (Crossref), https://doi.org/10.1016/j.jbiomech.2019.07.024.

Mummolo, Carlotta, et al. “Computational Evaluation of Load Carriage Effects on Gait Balance Stability.” Computer Methods in Biomechanics and Biomedical Engineering, vol. 19, no. 11, Aug. 2016, pp. 1127–36. Taylor and Francis+NEJM, https://doi.org/10.1080/10255842.2015.1110146.

O’Leary, Thomas J., et al. “Sex Differences in Neuromuscular Fatigability in Response to Load Carriage in the Field in British Army Recruits.” Journal of Science and Medicine in Sport, vol. 21, no. 6, June 2018, pp. 591–95. www.jsams.org, https://doi.org/10.1016/j.jsams.2017.10.018.

Oliveira, Camila Fonseca, et al. “Kinematic Changes during Prolonged Fast-Walking in Old and Young Adults.” Frontiers in Medicine, vol. 4, 2017. Frontiers, https://www.frontiersin.org/articles/10.3389/fmed.2017.00207.

Park, Huiju, et al. “Impact of Ballistic Body Armour and Load Carriage on Walking Patterns and Perceived Comfort.” Ergonomics, vol. 56, no. 7, July 2013, pp. 1167–79. Taylor and Francis+NEJM, https://doi.org/10.1080/00140139.2013.791377.

Paul, Sohini, et al. “Effect of Uphill Walking with Varying Grade and Speed during Load Carriage on Muscle Activity.” Ergonomics, vol. 59, no. 4, Apr. 2016, pp. 514–25. Taylor and Francis+NEJM, https://doi.org/10.1080/00140139.2015.1073792.

Power, Geoffrey A., et al. “Power Loss Is Greater Following Lengthening Contractions in Old versus Young Women.” Age (Dordrecht, Netherlands), vol. 34, no. 3, June 2012, pp. 737–50. PubMed, https://doi.org/10.1007/s11357-011-9263-z.

Reimann, Hendrik, et al. “Complementary Mechanisms for Upright Balance during Walking.” PLOS ONE, vol. 12, no. 2, Feb. 2017, p. e0172215. PLoS Journals, https://doi.org/10.1371/journal.pone.0172215.

Rice, Hannah, et al. “Influence of a 12.8-Km Military Load Carriage Activity on Lower Limb Gait Mechanics and Muscle Activity.” Ergonomics, vol. 60, no. 5, May 2017, pp. 649–56. Taylor and Francis+NEJM, https://doi.org/10.1080/00140139.2016.1206624.

Sawicki, Gregory S., and Daniel P. Ferris. "Mechanics and Energetics of Level Walking with Powered Ankle Exoskeletons." The Journal of Experimental Biology, vol. 211, no. Pt 9, May 2008, pp. 1402–13. PubMed, https://doi.org/10.1242/jeb.009241.

Seay, Joseph F., et al. "Lower Extremity Biomechanical Changes Associated With Symmetrical Torso Loading During Simulated Marching." Military Medicine, vol. 179, no. 1, Jan. 2014, pp. 85–91. Silverchair, https://doi.org/10.7205/MILMED-D-13-00090.

Silder, Amy, et al. "Men and Women Adopt Similar Walking Mechanics and Muscle Activation Patterns during Load Carriage." Journal of Biomechanics, vol. 46, no. 14, Sept. 2013, pp. 2522–28. DOI.org (Crossref), https://doi.org/10.1016/j.jbiomech.2013.06.020.

Simpson, Katrina M., et al. "Backpack Load Affects Lower Limb Muscle Activity Patterns of Female Hikers during Prolonged Load Carriage." Journal of Electromyography and Kinesiology, vol. 21, no. 5, Oct. 2011, pp. 782–88. ScienceDirect, https://doi.org/10.1016/j.jelekin.2011.05.012.

Sturdy, Jordan. DEVELOPMENT OF A MUSCULOSKELETAL MODEL FOR LOAD CARRIAGE. June 2019, p. 38.

Wallace, Jonathan W., et al. "Time-Dependent Neuromuscular Parameters in the Plantar Flexors Support Greater Fatigability of Old Compared with Younger Males." Experimental Gerontology, vol. 74, Feb. 2016, pp. 13–20. ScienceDirect, https://doi.org/10.1016/j.exger.2015.12.001.

Wills, Jodie A., et al. "Ankle and Knee Moment and Power Adaptations Are Elicited through Load Carriage Conditioning in Males." Journal of Biomechanics, vol. 97, Dec. 2019, p. 109341. ScienceDirect, https://doi.org/10.1016/j.jbiomech.2019.109341.

Zeni, J. A., et al. "Two Simple Methods for Determining Gait Events during Treadmill and Overground Walking Using Kinematic Data." Gait & Posture, vol. 27, no. 4, May 2008, pp. 710–14. ScienceDirect, https://doi.org/10.1016/j.gaitpost.2007.07.007.

Zhang, Li, et al. "Effect of Fatigue on Kinematics, Kinetics and Muscle Activities of Lower Limbs during Gait." Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine, July 2022. Sage UK:

London, England, journals.sagepub.com, https://doi.org/10.1177/09544119221112516.