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Gait and neuromuscular dynamics during level and uphill walking carrying military loads

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ABSTRACT
The neuromuscular system responds to perturbation and increasing locomotor task difficulty by altering the stability of neuromuscular output signals. The purpose of this study was to determine the effects of two different military load carriage systems on the dynamic stability of gait and muscle activation signals. 14 army office cadets (20 ± 1 years) performed 4-minute treadmill walking trials on level (0%) and uphill (10%) gradients while unloaded, and with 11 kg backpack and 11 kg webbing loads while the activity of 6 leg and trunk muscles and the motion of the centre of mass (COM) were recorded. Loaded and uphill walking decreased stability and increased magnitude of muscle activations compared to loaded and level gradient walking. Backpack loads increased the medio-lateral stability of COM and uphill walking decreased stability of vertical COM motion and increased stride time variability. However, there was no difference between the two load carriage systems for any variable. The reduced stability of muscle activations in loaded and uphill conditions indicates an impaired ability of the neuromuscular control systems to accommodate perturbations in these conditions which may have implications on the operational performance of military personnel. However, improved medio-lateral stability in backpack conditions may indicate that participants were able to compensate for the loads used in this study, despite the decreased vertical stability and increased stride time variability evident in uphill walking. This study did not find differences between load carriage systems however, specific load carriage system effects may be elicited by greater load carriage masses.

KEYWORDS
Military; load carriage; biomechanics; Lyapunov exponent; electromyography

Highlights

- Loaded and uphill walking decreased dynamic stability of muscle activations
- Lower activation stability indicates impaired neuromotor resistance to perturbation
- Backpack and webbing loads produced similar effects on muscle activations

Introduction

Load carriage is an operational requirement for military personnel (Knapik, Reynolds, & Harman, 2004; Walsh & Low, 2021). The added inertia of loads, most commonly carried on the trunk, results in alterations to walking gait (Liew, Morris, & Netto, 2016; Walsh & Low, 2021). Increases in lower limb joint moments (Quesada, Mengelkoch, Hale, & Simon, 2000; Rice, Fallowfield, Allsopp, & Dixon, 2017; Seay, Fellin, Sauer, Frykman, & Bensel, 2014), joint contact forces (Lenton et al., 2018) and muscle activity (Paul, Bhattacharyya, Chatterjee, & Majumdar, 2016; Rice et al., 2017; Sessoms et al., 2020), in addition to decreased trunk sway (Sessoms et al., 2020), trunk side flexion regularity (Morrison, Hale, & Brown, 2019) and hip, knee and ankle range of motion (Morrison et al., 2019; Rice et al., 2017; Seay et al., 2014) are commonly reported in soldiers when carrying loads. Load carriage has also been shown to decrease the stability of leg joint kinematics (Arellano, O'Connor, Layne, & Kurz, 2009) and medio-lateral trunk stability (Walsh, Low, & Arkesteijn, 2018) in young civilian and older adult populations. These gait alterations are the result of the mechanical demands of load carriage and attempt to safely accommodate the system perturbation caused by the load. However, in military personnel, stress fractures, overuse soft tissue injury, thoracic and low back pain, foot blisters and neuropathies are commonly reported in association with load

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carriage (Andersen, Grimshaw, Kelso, & Bentley, 2016; Cohen, Gallagher, Davis, Griffith, & Carragee, 2012; Knapik et al., 2004; Orr, Pope, Johnston, & Coyle, 2014, 2017). The high injury prevalence suggests that the neuromuscular and skeletal systems are often unable to sufficiently accommodate the perturbation caused by additional load.

Successfully accommodating the perturbation caused by carrying loads, particularly in environments with challenging terrains such as uneven ground and steep gradients, is essential for preventing injury and maintaining operational effectiveness (Walsh & Low, 2021). Increases in muscle activity are required to accommodate the added inertia of the load and produce sufficient propulsive forces. It has also been suggested, when unloaded, that the local dynamic stability of modular neuromuscular activations during locomotion decreases in response to accelerating speeds (Kibushi, Hagio, Moritani, & Kouzaki, 2018), but increases in running compared to walking and in response to external perturbations (Santuz et al., 2020). The stability of muscle activation patterns provides insight into the control strategies of the central nervous system (CNS), being indicative of the dynamical nature of descending motor commands (Santuz et al., 2020; Walsh, 2021). Therefore, understanding the stability of muscle activation will identify control mechanisms when carrying loads and walking on gradients.

The extent to which load may impact the mechanical and therefore neuromuscular function of walking gait is dependent on both the magnitude and distribution of the load (Birrell & Haslam, 2010; Lenton et al., 2019; Seay et al., 2014). There are a number of load carriage strategies adopted by militaries, such as traditional backpacks, weaponry held in the hands or on the trunk and webbing, a vest with load carried close to the waist (Knapik et al., 2004; Walsh & Low, 2021). Backpack load carriage systems, which position the mass behind the trunk, require compensations in trunk, hip and knee kinematics and kinetics to prevent a loss of stability (Liew et al., 2016). Whereas, systems that position the load closer to the hips or distributed around the waist, such as webbing systems, require less compensation in joint kinetics (Lenton et al., 2019; Seay et al., 2014). This indicates lower mechanical demands and potentially different neuromuscular control adaptions to the load. Understanding the neuromuscular control requirements of the different load carriage strategies adopted can guide selection and design of military load carriage systems and physical training requirements of military personnel.

The primary aim of this study was to determine the effect of two different military load carriage strategies on the neuromuscular and walking gait dynamic stability in level and uphill walking. The present study is the first to examine the effects of military load carriage on the dynamic stability of neuromuscular activity. This novel insight into the stability of muscle activity during loaded walking will extend our knowledge of the neuromuscular control of loaded walking and have potential implications for the training and use of load carriage in military personnel. It was hypothesised that carrying loads and walking uphill would decrease the stability of gait and muscle recruitment and that carrying a backpack would have a larger effect than carrying webbing. Furthermore, it was hypothesised that load carriage and uphill walking would increase spatio-temporal gait variability and the magnitude of muscle activations. Finally, it was hypothesised that load carriage effects would be greater walking uphill than walking on the level.

Methods

Participants

Fourteen Officer Cadets (*n* males: 8, *n* females: 6, age: 20 ± 1 years, body mass: 70.9 ± 8.6 kg, height: 1.72 ± 0.08 m), recruited from the Oxford University Officer Training Corp, a reserve unit of the British Army, volunteered to participate in this study. All participants had experience with the load carriage systems used in the study. Participants were excluded if they were not current army reservists, or presented musculoskeletal injury or neurological disorder. Participants were informed of the purpose of the study and provided written informed consent before commencing any of the data collection procedures. The study received ethical approval from the Oxford Brookes University Research Ethics Committee and procedures were conducted in accordance with the Declaration of Helsinki.

Procedures

All participants attended a single laboratory visit during which they performed treadmill walking trials in 3 load carriage conditions each in level (0% gradient) and uphill (10% gradient) walking, performing a total of 6 trials. The order that trials were completed was randomised. Two minutes of seated rest was provided between each trial. The 3 load carriage conditions consisted of an unloaded control condition, an 11 kg backpack (approximate dimensions: 60 × 40 × 15 cm) load and an 11 kg webbing load, consisting of a shoulder harness and a hip belt to carry pouches distributed around the waist. The load mass selected was lighter than loads commonly carried by military personnel and those previously reported to be required to elicit changes in spatio-temporal gait parameters (Liew et al., 2016; Walsh & Low, 2021). The webbing system is not commonly used to carry large masses and since the purpose of this study was to examine the effects caused by the different mechanical properties (i.e. distribution of load) of the systems, as opposed to the magnitude of the load carried, this was considered a reasonable compromise. In each of the load conditions the total mass was achieved by evenly distributing steel weights in the various compartments to avoid localised loads in a single section of the system.

Each treadmill trial lasted 4 min and required participants to walk at 1.7 $m{\cdot}s^{-1}$. This speed is representative of the standard average velocity of tabbing in the British Army, regardless of gradient, hence the same speed was utilised in each gradient condition despite the relatively greater difficulty of walking uphill. For each trial participants wore military boots, shorts and t-shirt. Before starting the experimental trials, participants performed 6 min of unloaded walking to warm up and familiarise with the treadmill.

Electromyographic (EMG) muscle activity was recorded using wireless EMG sensors with a 1 cm inter-electrode distance at 1000 Hz (Trigno Avanti, Delsys Inc., Natick, MA, USA) from the right upper Trapezius (TR), Erector Spinae (ES), Rectus Abdominis (RA), Biceps Femoris (BF), Rectus Femoris (RF) and Gastrocnemius Medialis (GM). Electrode sites were determined according to the SENIAM guidelines (Hermens et al., 1999). Prior to sensor attachment the skin surface was shaved and cleaned with an alcohol wipe. A 9-axis inertial measurement unit (IMU: Trigno Avanti, Delsys Inc., Natick, MA, USA) was placed over the L5 vertebra recording synchronously with the EMG sensors at 150 Hz, to measure the movements of the centre of mass (COM).

Gait dynamics

All data was analysed by custom written MATLAB (R2016b, Mathworks Inc., Natick MA, USA) scripts. Initial contact gait events were determined from the anterior-posterior (AP) acceleration of the IMU placed over L5. The AP acceleration signal was filtered twice separately with 20 and 2 Hz cut-off frequencies using second order dual-pass Butterworth filters. Initial contacts were determined as the points of the peaks in the 20 Hz filtered signal which immediately precede positive to negative zero-crossings in the 2 Hz filtered signal (McCamley, Donati, Grimpampi, & Mazzà, 2012). The middle 100 gait cycles were separated for the AP, medio-lateral (ML) and vertical (VT) axis acceleration signals. The average stride time (ST_{MEAN}) was determined as the average time between subsequent ipsilateral heel strikes. The stride time variability (ST_{VAR}) was determined as the standard deviation of the stride time. The data for the 100 strides were then interpolated to have 20,000 data points per axis.

From the interpolated COM acceleration signals the maximum Lyapunov exponent (MLE_{COM}) was calculated separately for each direction using the Rosenstein algorithm (Rosenstein, Collins, & De Luca, 1993). For the calculation of MLE_{COM} the state space of each acceleration signal was determined using the delay embedding method, with delay of 12 and embedding dimension of 5. Delay and embedding dimensions were determined to be sufficient for all participants and were calculated using the Average Mutual Information and False Nearest Neighbours algorithms, respectively (Dingwell & Cusumano, 2000; Raffalt, Senderling, & Stergiou, 2020). The nearest neighbour for each point in the state space was found as the points with the smallest Euclidean distance. Nearest neighbours with a temporal separation of less than the mean period of the signal were ignored (Rosenstein et al., 1993). The Euclidean distance of each pair of neighbours was followed for the length of the signal creating as many divergence curves as there were points in the state space. The MLE_{COM} was determined from the log average divergence curve as the slope of the linear period from the first point to the duration of 1 stride. MLE represents the local dynamic stability of a system, the system's ability to resist small internal perturbations, as the exponential rate of divergence of points in close proximity in the state space of the signal. A larger MLE value indicates a less stable system as the rate of divergence of points in state space is greater.

Muscle activity dynamics

The EMG data for the middle 100 strides were separated and the signal of each muscle was bandpass filtered using a second order dual-pass Butterworth filter with 50 and 450 Hz cut-off frequencies. Bandpass filtered data were demeaned, full-wave rectified and lowpass filtered with a second order dual-pass Butterworth filter with a 15 Hz cut-off frequency to create the linear envelope for each muscle. The envelope of each muscle was normalised to the maximum value for each muscle in the control, 0% gradient trial. To quantify the magnitude of activation the mean of the normalised EMG signal for each muscle was calculated (EMG_{MEAN}).

To quantify the stability of neuromuscular control signals the maximum Lyapunov exponent of muscle activation (MLE_{EMG}) patterns was determined. A state-space was created from the muscle activation signals of each muscle and the first differential of each muscle activation signal producing a 12-dimensional state space, which is sufficient to quantify the dynamics of muscle activity in walking (Kang & Dingwell, 2009). This approach also has the advantage of avoiding the need to calculate an appropriate delay and embedding dimension which is very sensitive to the presence of noise in biological signals (Santuz et al., 2020). From the muscle activity state space, the MLE_{EMG} was calculated as described for MLE_{COM}.

Statistics

Data were tested for normality using the Shapiro–Wilk test. For all EMG variables a 3 (load condition) × 2 (gradient) two-way repeated measures MANOVA was performed to determine the effect of load condition and gradient and separate 3 (load condition) × 2 (gradient) two-way repeated measures MANOVA was performed for all gait variables. The Wilk's lambda test statistic was used for both MANOVA. For significant multivariate effects, univariate 3 (load condition) × 2 (gradient) two-way repeated measures ANOVA were performed. Significant main effects of load condition were explored with post hoc pairwise comparisons with a Bonferroni correction. The 95% confidence intervals (CI) were provided for pairwise comparisons. For effects of load and interaction effects which violated the assumption of sphericity, determined using Mauchly's test, the Greenhouse-Geisser correction of the degrees of freedom was used. For all main and interaction effects the partial eta squared (η_p^2) effect size was calculated. Values of 0.01, 0.06 and 0.14 represent small, moderate and large effects, respectively. For all tests the a priori level of significance was $p < 0.05$. All statistical analysis was performed using SPSS (v26, IBM Corp., NY USA).

Results

Data for gait MLE_{COM} variables are shown in Figure 1 and for spatio-temporal variables are shown in Figure 2. There were significant multivariate effects of load ($\lambda = 0.31$, $F(10,44) = 3.51$, $p = 0.002$, $\eta_p^2 = 0.44$) and gradient ($\lambda = 0.15$, $F(5,9) = 10.63$, $p = 0.001$, $\eta_p^2 = 0.86$) on gait dynamics variables. However, there was no multivariate interaction effect ($\lambda = 0.50$, $F(10,44) = 1.80$, $p = 0.088$, $\eta_p^2 = 0.29$).

Significant univariate effects of load were found on ML MLE_{COM} ($F(2,26) = 9.38$, $p < 0.001$, $\eta_p^2 = 0.42$). However, there were no univariate load effects on any other variable. ML MLE_{COM} was significantly lower when carrying a backpack when compared to the control condition ($p < 0.001$, CI: 0.13, 0.37). There were also significant univariate effects of gradient on VT MLE_{COM} ($F(1,13) = 21.64$, $p < 0.001$, $\eta_p^2 = 0.63$), ST_{MEAN} ($F(1,13) = 8.37$, $p = 0.013$, $\eta_p^2 = 0.39$) and ST_{VAR} ($F(1,13) = 10.37$, $p = 0.007$, $\eta_p^2 = 0.44$). However, there were no effects on other gait variables. VT MLE_{COM} and ST_{VAR} were greater during uphill than level walking and ST_{MEAN} was greater during level than uphill walking.

Data for MLE_{EMG} is shown in Figure 3 and for all EMG_{MEAN} are shown in Figure 4. There were significant multivariate effects of load ($\lambda = 0.06$, $F(14,40) = 8.91$, $p < 0.001$, $\eta_p^2 = 0.76$) and gradient ($\lambda = 0.07$, $F(7,7) = 13.95$, $p = 0.001$, $\eta_p^2 = 0.93$) on gait dynamics variables. However, there was no multivariate interaction effect ($\lambda = 0.40$, $F(14,40) = 1.65$, $p = 0.108$, $\eta_p^2 = 0.37$).

There were significant univariate load condition effects on MLE_{EMG} ($F(2,26) = 10.48$, $p < 0.001$, $\eta_p^2 = 0.45$) and on EMG_{MEAN} on TRAP ($F(2,26) = 11.54$, $p < 0.001$, $\eta_p^2 = 0.47$), ES ($F(2,26) = 10.17$, $p < 0.001$, $\eta_p^2 = 0.44$), RF ($F(2,26) = 1.62$, $p < 0.001$, $\eta_p^2 = 0.51$) and GM ($F(2,26) = 40.05$, $p < 0.001$, $\eta_p^2 = 0.76$) but with no effect for any other muscle. The MLE_{EMG} was greater in webbing ($p = 0.006$, CI: −0.19, −0.03) and backpack ($p = 0.002$, CI: −0.16, −0.04) load conditions than control. The EMG_{MEAN} of TRAP (webbing: $p = 0.007$, CI: −19.87, −3.18; backpack: $p = 0.007$, CI: −11.73, −1.81), RF (webbing: $p = 0.033$, CI: −9.87, −0.38; backpack: $p < 0.001$, CI: −9.71, −4.64) and GM (webbing: $p < 0.001$, CI: −8.44, −4.47; backpack: $p < 0.001$, CI: −8.51, −4.64) was greater in webbing and backpack load conditions than control and ES was greater in webbing than control ($p = 0.002$, CI: −6.52, −1.56). However, there were no differences between backpack and webbing conditions. The MLE_{EMG} ($F(1,13) = 16.45$, $p = 0.001$, $\eta_p^2 = 0.56$), RF ($F(1,13) = 16.14$, $p = 0.001$, $\eta_p^2 = 0.55$) and GM ($F(1,13) = 89.94$, $p < 0.001$, $\eta_p^2 = 0.87$) were also greater in uphill than level walking.

Discussion

This study aimed to determine the effect of military load carriage and gradient during treadmill walking on gait and muscle activity dynamics. It was found that carrying loads of 11 kg resulted in less dynamically stable muscle activity and greater activity in the TRAP, ES, RF and GM, in conjunction with more stable ML COM motion with the backpack load. However, this study found no difference in backpack and webbing loads for any variable. Walking uphill also resulted in lower stability of muscle activations and greater activity of the RF and GM with less stable VT COM motion but greater ST_{VAR} and shorter ST_{MEAN}.

The lower dynamic stability (i.e. greater MLE_{EMG}) of muscle activations during loaded and uphill walking

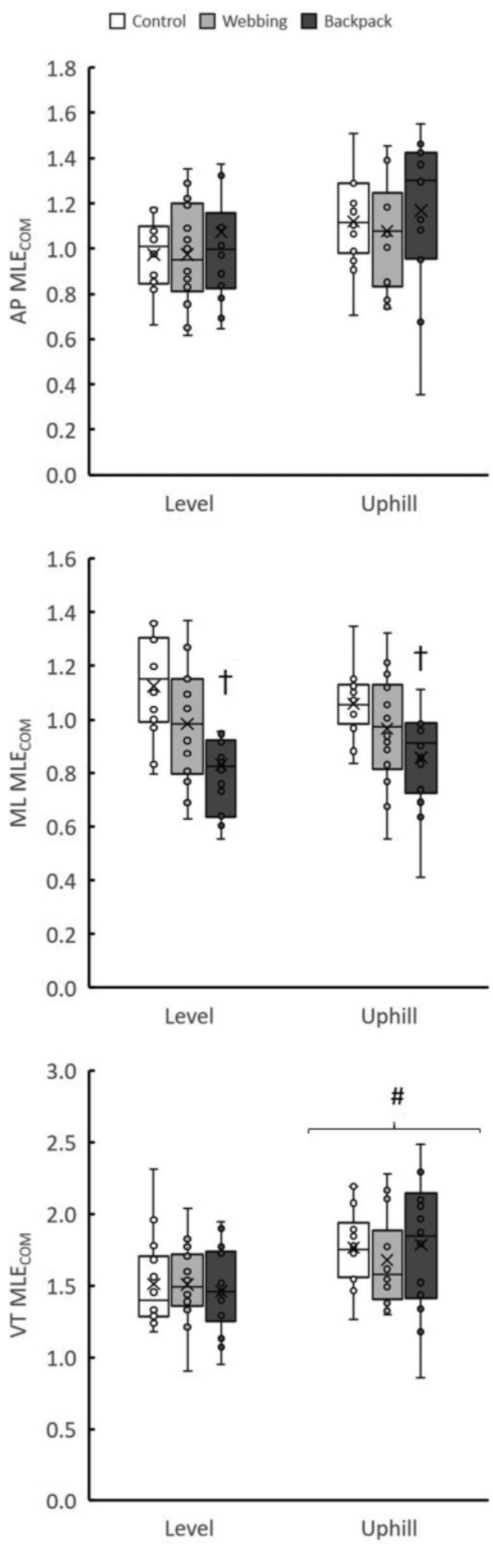

Figure 1. Boxplots for the anterior-posterior (AP), medio-lateral (ML) and vertical (VT) centre of mass maximum Lyapunov exponent (MLE_{COM}) in all load and gradient conditions. The mean of each condition is represented by an X. The upper, middle and bottom horizontal line of each box represent the 1st quartile, median and 3rd quartile of the data, respectively, and the error bars indicate the minimum and maximum values. †indicates that the value is lower than the unloaded condition. #indicates that the value is greater than the level gradient condition.

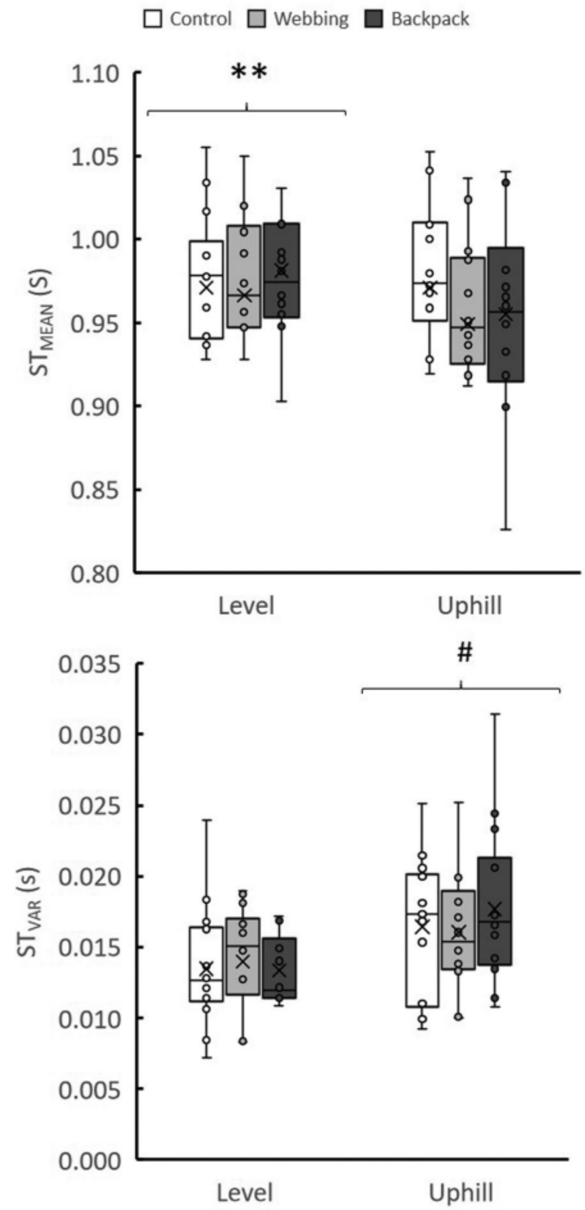

Figure 2. Boxplots for the average stride time (ST_{MEAN}) and stride time variability (ST_{VAR}) variables in all load and gradient conditions. The mean of each condition is represented by an X. The upper, middle and bottom horizontal line of each box represent the 1st quartile, median and 3rd quartile of the data, respectively, and the error bars indicate the minimum and maximum values. ** indicates that the value is greater than the uphill gradient condition. # indicates that the value is greater than the level gradient condition.

provide a novel insight into the effects of load carriage on neuromuscular control of gait. These findings are in agreement with previous research which has found that the stability of muscle synergy signals decreases with increasing task difficult, such as increased walking speeds (Kibushi et al., 2018; Walsh, 2021) and greater loads and rate in a lifting task (Graham & Brown, 2014). Decreased muscle activation stability when carrying

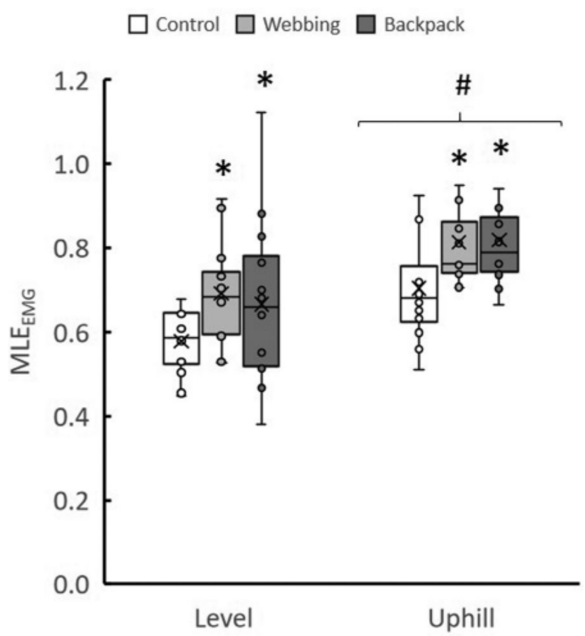

Figure 3. Boxplots for the EMG maximum Lyapunov exponent (MLE_{EMG}) in all load and gradient conditions. The mean of each condition is represented by an X. The upper, middle and bottom horizontal line of each box represent the 1st quartile, median and 3rd quartile of the data, respectively, and the error bars indicate the minimum and maximum values. * indicates that the value is greater than the unloaded condition. # indicates that the value is greater than the level gradient condition.

loads and walking uphill may indicate that the CNS was not able to maintain the stability of neuromuscular output when faced with the greater task difficulty in each condition compared to unloaded, level walking. This finding is of note for military personnel as the decreased neuromuscular stability may indicate that the CNS control systems are less resistant to perturbation when carrying loads. This would have implications in the field where soldiers are often exposed to challenging and unpredictable terrain. Consequently, the ability to navigate this terrain safely could be impeded and reduced neuromuscular stability may explain the reported decrease in agility, obstacle navigation and marksmanship when carrying loads (Jaworski, Jensen, Niederberger, Congalton, & Kelly, 2015; Joseph, Wiley, Orr, Schram, & Dawes, 2018; O'Neal, Hornsby, & Kelleran, 2014). The findings of this study highlight the importance of training in various terrains and operational skills when carrying loads. Exposing the neuromuscular system to the combined challenges of load carriage and operational tasks regularly may prevent adversely large loss of neuromuscular stability in the field.

An alternative interpretation of the current results for the stability of muscle activations is that in the current population of healthy adults, experienced in load carriage, there may have been sufficient capacity in the neuromuscular system to accommodate the greater task difficulty. Therefore, the CNS was not required to increase the stability of neuromuscular signals to prevent a large decrease in gait stability. This redundancy in the neuromuscular system may explain the decrease in stability of muscle activations as greater activity is required to accommodate the demands of load carriage (Kibushi et al., 2018). In contrast, it has previously been demonstrated that running and walking with external perturbations had more stable muscle activation synergies than unperturbed walking (Santuz et al., 2020). It could be that running and perturbed walking represent more challenging tasks than loaded and uphill walking requiring more stable muscle activations.

Interestingly, despite the decreased stability of muscle activations in loaded conditions, the only change in the stability of COM movements was found in the ML direction where stability was greater (i.e. lower MLE_{COM}) in the backpack condition compared to the unloaded condition. Similarly to this study, it has previously been found that loads carried around the hips, like that of the webbing condition, had no effect on AP COM stability (Arellano, Layne, O'Connor, Scott-Pandorf, & Kurz, 2009), despite decreases in stability of the joint kinematics (Arellano et al., 2009). This suggests that participants were able to compensate for the added mass without decrement to COM stability in the AP and VT directions. It may also be the case that due to the mechanical constraints of walking, stability in these directions are minimally affected by load carriage (Arellano et al., 2009). It is reasonable to assume that the backpack, with the load mass distributed further from the COM, required a greater compensation than the webbing load, particularly in the ML direction where gait stability is actively controlled (Donelan, Shipman, Kram, & Kuo, 2004; Mahaki, Bruijn, & Van Dieën, 2019). Therefore, with the relatively light load used in this study compensations for the perturbation caused by the load resulted in increased stability in the ML direction. However, contrary to the present study, weighted vest loads of 40% body mass reduced the ML stability of COM in level walking (Ignasiak et al., 2019), possibly indicating heavier loads exceed a threshold below which loads can be accommodated without a loss in stability. Additionally, the absence of load effects on ST_{MEAN} and ST_{VAR} with the 11 kg load carried in this study are in agreement with previous systematic reviews where minimal load effects on spatio-temporal gait parameters were reported (Walsh & Low, 2021) and changes in cadence were only found with heavier loads above 30% body mass (Liew et al., 2016).

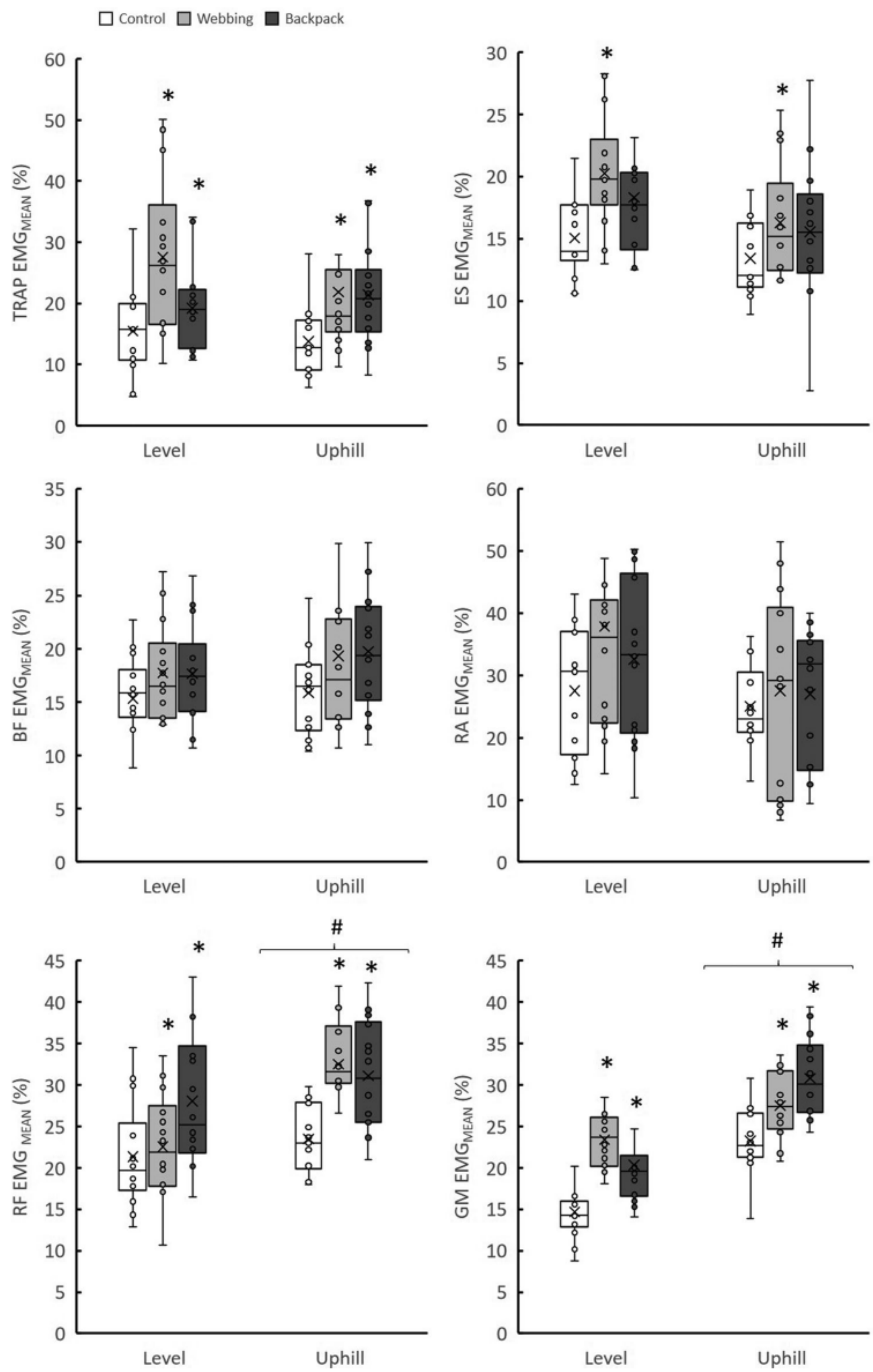

Figure 4. Boxplots for the average EMG activity (EMG_{MEAN}) of the Trapezius (TRAP), Erector Spinae (ES), Biceps Femoris (BF), Rectus Abdominis (RA), Rectus Femoris (RF) and Gastrocnemius Medialis (GM) in all load and gradient conditions. The mean of each condition is represented by an X. The upper, middle and bottom horizontal line of each box represent the 1st quartile, median and 3rd quartile of the data, respectively, and the error bars indicate the minimum and maximum values. * indicates that the value is greater than the unloaded condition. # indicates that the value is greater than the level gradient condition.

In addition to the load effects on gait found in this study, a lower ST_{MEAN} and greater ST_{VAR} were found in uphill walking than level walking, indicating that when walking uphill participants took shorter, more variable steps, similar to previous findings (Fellin, Seay, Gregorczyk, & Hasselquist, 2016; Kimel-Naor, Gottlieb, & Plotnik, 2017; Vickery-Howe, Clarke, Drain, Dascombe, & Middleton, 2020). The higher ST_{VAR} associated with uphill walking suggests the control of the gait pattern was compromised in these conditions. Similarly, the dynamic stability in the VT direction was reduced (i.e. greater MLE_{COM}) in uphill walking, further indicating impaired neuromotor control when walking uphill. However, there was no effect of gradient on the AP or ML stability. A possible interpretation for this finding may be that participants preferentially controlled stability in these directions whilst allowing the VT stability to decrease. An alternative interpretation is that the added demand for vertical propulsion in uphill walking presents a greater challenge to the control systems in this direction. This is reflected in the greater EMG_{MEAN} in uphill walking compared to level walking of the RF and GM, muscles which are primarily responsible for vertical and forward propulsion.

Greater EMG_{MEAN} was also found in both loaded conditions compared to unloaded walking for the TRAP, RF and GM and in webbing for ES. These findings are in accordance with previous findings (Lindner et al., 2012; Paul et al., 2016; Rice et al., 2017; Sessoms et al., 2020). Greater activity is indicative of the need to overcome the added inertia for vertical and forward propulsion (i.e. RF and GM activity), additional support for the trunk (i.e. ES activity) and to support the shoulder girdle (i.e. TRAP activity). The greater EMG_{MEAN} of multiple muscles in the loaded walking conditions indicates the greater neuromuscular demand in these conditions. This greater demand may explain the decreased dynamic stability of muscle activations as the CNS attempts to accommodate the additional demand. The EMG_{MEAN} of ES was only greater in webbing but not backpack conditions. A large inter-subject variability was found in ES activity in backpack conditions. It is possible that despite efforts to avoid interference of the load on the EMG recording and subsequent visual inspection of the data that this variability was the result of artefacts caused by the load. Contrary to the expected outcomes, RA activity was not increased by load carriage. This was likely the result of the substantial inter-subject variability demonstrated for this muscle.

Contrary to our hypothesis, no difference was found between the two load conditions and no interaction effects were present. However, backpack loads had an effect on ML MLE_{COM} when compared with unloaded walking that was not found in the webbing condition. It was anticipated that the webbing condition would have a smaller effect on neuromuscular and gait stability as the load is positioned closer to the centre of mass than backpack loads. However, the findings of this study suggest that carrying loads about the hips does not offer an advantage on gait or neuromuscular stability compared to backpack loads. It is possible that greater load mass would have elicited differences between conditions as the ability of participants to compensate for the added mass is reduced, particularly as the loads carried by military personnel are generally larger than that used in this study. However, the load mass utilised in this study was not intended to specifically replicate military operational loads but to provide a comparison between military specific load systems carried on the back and with the load distributed around the torso. Since the webbing load carriage system is not designed to carry significant mass, a load mass was selected that would be reasonable to carry in the webbing but be sufficient to cause an effect. It is possible that a heavier load mass would have provided additional insight into potential differences in gait and neuromuscular stability between the two load carriage systems.

There were limitations of this study that should be considered. Firstly, only a single load mass was utilised in this study, which was lower than those general carried by military personnel in the field. Additionally, in both conditions participants were required to walk on a fixed paced treadmill which may have prevented participants adopting their preferred gait strategy. However, under marching conditions soldiers would be expected to maintain these average walking speeds when loaded. Treadmill walking also allows for the analysis of continuous walking that would be challenging to achieve in more ecologically valid conditions. Finally, the study was conducted on military reserves, it is feasible that those who are exposed more regularly to heavy load carriage in a variety of operational settings would have a different response to the participants in this study. Future work should examine the effects of heavier loads in a variety of load carriage systems on the neuromuscular stability of military personnel with a particular focus on the relationship with agility and the ability to navigate challenging terrains.

Conclusion

The present study provides a novel understanding of the effect of military load carriage and gradient on the stability of muscle activations when walking at operationally relevant speeds. In both loaded and uphill walking the stability of muscle activations is reduced, indicating an impaired ability of the neuromuscular control systems to accommodate perturbations in these conditions which may have implications on the operational performance of military personnel. The reduced stability of neuromuscular output likely stems from the increased demand caused by loaded and uphill walking, as indicated by increased muscle activity when compared to control conditions. Despite decreased neuromuscular output stability, the ML stability of the COM increased when carrying backpack loads possibly as a result of compensatory control adjustments possible with the relatively light load carried. Although the mechanical properties of the two load carriage systems utilised in this study are different there was no difference between load carriage systems or interaction effects for any variable. Differences between load carriage systems may have been elicited by greater load carriage masses. However, uphill walking resulted in greater gait variability and reduced VT COM stability, indicating impaired gait control when compared to level walking.

Disclosure statement

No potential conflict of interest was reported by the author(s).

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