Stable and unstable load carriage effects on the postural control of older adults.

Gregory S Walsh[1], Daniel C Low[2,3], Marco Arkesteijn[2]

[1]Department of Sport, Health Sciences and Social Work, Oxford Brookes University, Oxford, UK, OX23 0BP.

[2]Institute of Biological, Environmental and Rural Sciences, Aberystwyth University, Aberystwyth, UK, SY23 3FD.

[3]Centre for Human Performance, Exercise and Rehabilitation, Brunel University London, London, UK.

Corresponding author: Gregory S Walsh

Email: gwalsh@brookes.ac.uk, alternate email: greg_s_walsh@yahoo.co.uk, Tel: +44 (0) 1865 483704

Department of Sport, Health Sciences and Social Work, Oxford Brookes University, Oxford, UK, OX23 0BP.

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Abstract

The aim of this study was to investigate the effects of backpack load carriage on quiet standing postural control and limits of stability of older adults. Fourteen older adults (65±6 years) performed quiet standing and a forward, right and left limits of stability test in 3 conditions, unloaded, stable and unstable backpack loads while activity of 4 leg muscles was recorded. Stable and unstable loads decreased postural sway (main effect η_p^2=0.84, stable: *p*<.001, unstable: *p*<.001), medio-lateral (main effect η_p^2=0.49, stable: *p*=.002, unstable: *p*=.018) and anterior-posterior (main effect η_p^2=0.64, stable: *p*<.001, unstable: *p*=.001) fractal dimension and limits of stability distance (main effect η_p^2=0.18, stable: *p*=.011, unstable: *p*=.046) compared to unloaded. Rectus Femoris (main effect η_p^2=0.39, stable: *p*=.001, unstable: *p*=.010) and Gastrocnemius (main effect η_p^2=0.30, unstable: *p*=.027) activity increased in loaded conditions during limits of stability and quiet standing. Gastrocnemius-Tibialis Anterior coactivation was greater in unstable load than stable loaded quiet standing (main effect η_p^2=0.24, *p*=.040). These findings suggest older adults adopt conservative postural control strategies minimising the need for postural corrections in loaded conditions. Reduced limits of stability may also increase fall risk when carrying a load. However, there was no difference between unstable and stable loads for postural control variables.

Keywords: Quiet standing, Limits of stability, Fractal dimension, Electromyography, Load carriage

Word Count: 3715

Introduction

Disturbances to the postural control system can come from numerous sources including physical perturbations, muscle fatigue and load carriage[1–3]. It was demonstrated previously that a period of prolonged walking can lead to postural control alterations in older adults[4]. A potential explanation is that the fatigue results in acute changes to the force production capabilities of a muscle resulting in a smaller muscle force production to body mass ratio. This observation was supported by Ledin et al.,[1] although, they also found that load carriage had a larger impact on postural control than muscle fatigue. Carrying a load on the trunk, e.g. wearing a backpack, artificially increases the mass of the trunk. This negatively impacts the ability to perform postural corrections as the force output needed for postural corrections is increased[5].

Previous studies investigating the effect of load carriage on postural control of younger adults found increased postural sway[1,5–7] and complexity of postural sway[5]. During tasks requiring participants to move the centre of mass (COM) towards the limits of stability, handheld loads reduce the maximum distance young adults can move the COM. Together these findings suggest that load carriage reduces postural stability[5] which could have implications for fall risk in older adults.

Unstable loads have different effects on postural control than stable loads[3], suggesting the type of load can also impact postural control. An unstable load held in the hands increases sway velocity and area in young adults[3]. In addition, older adults are more likely to be affected by load carriage[8,9]. Movements of an unstable load require individuals to produce additional corrective forces to attenuate perturbation provided by the load, increasing the demand on the postural control system. The use of perturbations to investigate the stability of the postural control system is common[10,11]. Load carriage perturbs the postural control system by increasing the mass that must be supported and controlled[1,5,12], this effect can be magnified by unstable loads[3], providing insight into the mechanisms of postural control adopted by older adults when perturbed. Non-linear measures of postural sway complexity, such as the fractal dimension, can elucidate the neuromuscular control mechanisms[13,14] adopted when the system is perturbed.

Previous studies investigating the effect of load carriage on muscle activation in young adults have focussed either on muscles of the trunk and upper leg[3,12,15]. However, these studies have not investigated the activation of Triceps Surae muscles which are largely responsible for postural control[16,17]. In addition, it has been suggested that older adults utilise greater coactivation for postural control to compensate for age related neuromuscular decline[18]. Older adults may therefore rely on increased coactivation in response to added load.

It is currently unknown how load carriage affects older adults postural control, the limits of stability and muscle activation. Load carriage is a common task for community dwelling older adults and also provides a perturbation to the postural control system, therefore allowing the study of the robustness of the postural control system to perturbations in a commonly encountered paradigm[19]. The ability to respond to postural perturbations is essential for minimising the risk of falls in older adults[20]. To further explore the effect of perturbations the current study included an unstable loaded condition. The aim of this study was to determine the effect of stable and unstable load carriage on postural control, muscle activation and coactivation during quiet standing and limits of stability tests in older adults. It was hypothesised that stable and unstable load carriage would result in increased postural sway magnitude and complexity, with concurrent increases in lower limb muscle activity and coactivation. Additionally, it was hypothesised that stable and unstable loads would result in decreased limits of stability length and increased variability, with a concurrent increase in the lower limb muscle activity and coactivation. Finally, it was hypothesised that unstable loads would have a greater effect on postural control, muscle activation and coactivation than stable loads.

Methods

Participants: Fourteen community-dwelling older adults (n-females: 7, n-males: 7, age: 65±6 years, height: 1.70±0.10 m, mass: 74.0±13.0 kg, BMI: 25±3 $kg \cdot m^{-2}$) participated in this study. Participants were excluded if they suffered from neurological conditions such as stroke, Parkinson's disease or dementia. Exclusion criteria also included visual impairment or lower limb conditions that prevented walking or unaided quiet stance. The study received institutional ethical approval and all procedures were conducted according to the Declaration of Helsinki. All participants gave written informed consent, were aware of the nature of the study and were free to withdraw at any time.

Procedures: The postural control of participants was assessed during quiet standing and limits of stability (LOS), the ability to shift the COM toward the boundary of the base of support (BOS). Each assessment was completed under 3 load conditions; unloaded, stable load and unstable load, during a single visit. Both the stable and unstable loads were carried using a backpack with a chest strap and were equivalent to 15% of the participants' body mass (BM), to the nearest 0.1 kg[21]. In the stable and unstable load conditions 3 water-tight containers, with a volume of 3.6 litres each, were placed inside the backpack (Figure 1). For the stable load, steel weights in denominations of 0.1, 0.5 and 1 kg, were secured to the sides of the containers to mimic the COM of the unstable load and to prevent movement, and were evenly distributed between the 3 containers. To form the unstable load a volume of water equivalent to a mass of 7.5% of the participants BM was distributed evenly between the 3 containers and steel weights were then added to make up the total mass of the backpack to

15% of the participants BM. The order in which load conditions were performed was randomised across participants.

[Figure 1 here]

Postural control during quiet standing and LOS were performed with participants stood barefoot in a comfortable position on a force plate recording at 48 Hz (Kistler Instruments Ltd, Winterthur, Switzerland) with eyes open. The foot position of each participant was marked on a clear covering placed over the surface of the force plate to ensure the same position was adopted for each trial, as foot placement can alter the calculated postural sway parameters[22].

To assess quiet standing postural control, participants performed 5 trials of 60 seconds in each load condition. To test the LOS participants performed a total of 9, 30 second, trials in each condition. Each LOS trial consisted of 3 phases (Figure 2a). In phase 1 participants stood quietly for 10 seconds at which point they were asked to lean forward, right or left. Phase 2 began at the start of the lean movement and ended when participants reached a lean position they perceived as maximum distance that they could maintain without falling. The leaning movement was executed at a self-selected speed using an ankle strategy, whilst avoiding bending at the hips and knee, and keeping feet flat on the force plate surface. Trials in which participants visibly flexed the hips or knees, or lifted their heels were repeated. In phase 3, participants were asked to maintain the maximal lean position for the remainder of the 30 second trial. Three trials were performed for each lean direction.

[Figure 2 here]

During each quiet standing and LOS trial participants were fitted with reusable bipolar electrodes with a 2 cm inter-electrode distance (SX230-1000, Biometrics Ltd, UK) to measure the electromyographic (EMG) activity of the left Rectus Femoris (RF), Biceps Femoris (BF), Tibialis Anterior (TA) and Gastrocnemius Medialis (GM). A reference electrode was placed over the left radial head. Specific electrode placements are outlined in Table 1. The skin was prepared by shaving the area and cleaning with an alcohol wipe. The electrodes were attached to an 8-channel amplifier (range: ±4mV, gain: 1000, impedance: 1MΩ - K800, Biometrics Ltd, UK) before being A/D converted (CA-1000, National Instruments Corp., UK).

[Table 1 here]

Data Analysis: All quiet standing centre of pressure, LOS and muscle activation data analysis was performed using custom written MATLAB programmes (R2016a, Mathworks Inc., MA, USA).

Quiet Standing: The recorded centre of pressure (COP) signals were not filtered to avoid removing the natural variability of the signal which would impact the non-linear analyses as the complexity of the signal is removed[13]. The postural sway path length ($SWAY_{PL}$) was calculated as the resultant path length of the medio-lateral (ML) and anterio-posterior (AP) COP components. Fractal dimension (D_f) was calculated using Higuchi's algorithm[23] to estimate the complexity of the COP signals in the AP and ML directions. The time series $x=x(1),x(2),x(3),\ldots,x(N)$ is reconstructed into k new time series, $x(m,k)$ with initial time value m, and discrete time interval k:

$$x(m,k) = x(m), x(m+k), x(m+2k), \ldots, x\left(m + \left\lfloor \frac{N-m}{k} \right\rfloor k\right)$$

$$m = 1,2,3,\ldots,k$$

where N is the total number of samples. The maximum value of k (k_{max}) was predetermined as the point where a plot of k vs. D_f for increasing values of k plateaued. For the present study k_{max} values of 70 and 50 were selected for the AP and ML directions respectively. The average length ($L_m(k)$) of each new time series is calculated by:

$$L_m(k) = \frac{\sum_{i=1}^{\lfloor (N-m)/k \rfloor} |x(m+ik) - x(m+(i-1)k| (n-1)}{\lfloor (N-m)/k \rfloor k}$$

The average length for all signals with same k is then calculated as the mean of the lengths $L_m(k)$ for $m = 1, \ldots, k$. This process is repeated for each value of k in the range of 1-k_{max} resulting in the sum of average lengths ($L(k)$) for each k:

$$L(k) = \sum_{m=1}^{k} L_m(k)$$

The D_f is then determined as the slope of a linear least squares fit of the curve for ln($L(k)$) vs. ln(1/k).

Limits of Stability: The start and end of each phase during LOS trials was determined as the intersection points of separate linear least squares models fitted to the 3 distinct regions of the COP signal using the Shape Language Modelling MATLAB toolbox (R2016a, Mathworks Inc., MA, USA). The anterior-posterior, left-right boundaries of the base of support (BOS) were determined from the outline of the feet drawn on the force plate as the maximum displacement in each direction respectively. The length of the AP and ML BOS were then calculated as the distance between the anterior and posterior, and left and right boundaries.

The distance leaned in each LOS trial was calculated as the absolute distance between the average COP positions in phases 1 and 3 (Figure 2b). The distance leaned was reported relative to the total BOS length (LOS_{REL}) as a percentage in the AP direction for forward leaning trials and the ML direction for left and right leaning trials. A larger LOS_{REL} indicates a greater LOS and therefore better postural stability. The root mean square (LOS_{RMS}) was calculated from the detrended COP signal in phase 3 to indicate the variability of movement in the sustained period of leaning:

$$LOS_{RMS} = \sqrt{\frac{1}{N}\sum_{n=1}^{N} |COPn|^2}$$

where N is the length of the signal and COPn is the nth element of the COP signal.

Muscle Activation and Coactivation: Raw EMG signals were band-pass filtered with a dual-pass 2nd order Butterworth filter with 20-450 Hz cut-off frequencies before being full-wave rectified and low-pass filtered with a dual-pass 2nd order Butterworth filter with a 10 Hz cut-off frequency. Low-pass filtered EMG signals were normalised as a percentage of the maximum activity recorded during 60 seconds of unloaded quiet standing[24]. The average activity of each muscle (EMG_{MEAN}) was calculated for each quiet standing and LOS trial from the normalised signal.

The coactivation indices[25] (CI) of 2 muscle pairs (RF-BF and GM-TA) were calculated as follows:

$$CI = \frac{2Iant}{Itot} \text{ x } 100$$

Where *Itot* is the sum of the integrals of both muscles:

$$Itot = \int_{t1}^{t2} \left[EMG_{agonist} + EMG_{antagonist}\right](t)dt$$

and *Iant* is the total integral of antagonistic activity, defined as the muscle with the lower activity at each time point:

$$Iant = \int_{t1}^{t2} EMG1(t)dt + \int_{t2}^{t3} EMG2(t)dt$$

Where t1 and t2 denote periods that the activity of the first muscle of each pair is less than the second, and t2 and t3 denote the periods that the activity of second muscle is less than the first. Coactivation indices are expressed as a percentage of antagonistic activity with respect to total activity for each pair.

Statistics: All data were tested for normality using the Shapiro-Wilk test and for any data that violated the assumption of sphericity the Greenhouse-Geiser correction was applied. One-way repeated measures ANOVA were performed to determine the effect of load on quiet standing postural control variables ($SWAY_{PL}$, ML D_f and AP D_f) and muscle activation (RF, BF, GM and TA, and RF-BF and GM-TA CI). To determine the effects of load condition, direction and load x direction interaction effects on LOS variables (LOS_{REL} and LOS_{RMS}) and muscle activation (RF, BF, GM and TA and RF-BF, and GM-TA CI) two-way repeated measures ANOVA were performed. Post hoc pairwise comparisons with a Bonferroni correction were performed for significant main effects. Simple main effects with Bonferroni correction were used to explore significant interactions. For all tests α=0.05 and partial eta squared (η_p^2) was used as an estimate of effect size, values of 0.01, 0.06 and 0.14 were interpreted as small medium and large effects respectively[26]. All statistical analysis was performed using SPSS software (v22, IBM UK Ltd., Portsmouth, UK).

Results

There were significant load effects for $SWAY_{PL}$ (F(2,26)=68.75, *p*<.001, η_p^2=0.84), ML D_f (F(2,26)=12.61, *p*<.001, η_p^2=0.49) and AP D_f (F(2,26)=23.13, *p*<.001, η_p^2=0.64). All quiet standing variables were greater in unloaded compared to stable ($SWAY_{PL}$: *p*<.001, ML D_f: *p*=.002 and AP D_f: *p*<.001) and unstable ($SWAY_{PL}$: *p*<.001, ML D_f: *p*=.018 and AP D_f: *p*=.001) conditions. There were no differences between stable and unstable conditions (Table 2).

[Table 2 here]

There were also load effects for RF-BF (F(2,26)=3.74, *p*=.037, η_p^2=0.22) and GM-TA (F(2,26)=4.17, *p*=.027, η_p^2=0.24) CI. The RF-BF CI was lower in unstable than unloaded (*p*=.047), however GM-TA CI was greater in unstable than stable (*p*=.040) but there was no difference to unloaded (Figure 3). In addition, there was a load effect for GM EMG_{MEAN} (F(2,26)=5.48, *p*=.010, η_p^2=0.30) as unstable was greater than unloaded (*p*=.027). There were no load effects for any other muscle.

[Figure 3 here]

There was an effect of load for LOS_{REL} ($F(2,26)=2.77$, $p=.041$, $\eta_p^2=0.18$), the LOS_{REL} was greater in unloaded than stable (p=.011) and unstable ($p=.046$), however there was no load effect for LOS_{RMS} and no difference between stable and unstable (Figure 4). There were no effects of direction on LOS variables or interaction effects.

[Figure 4 here]

There was a load effect on RF EMG_{MEAN} ($F(1.4,18.2)=8.22$, $p=.006$, $\eta_p^2=0.39$) which was greater in stable ($p=.001$) and unstable ($p=.010$) than unloaded but no effects of direction for any muscle (Table 3). There was also an interaction effect for TA EMG_{MEAN} ($F(2.5,32.5)=3.77$, $p=.026$, $\eta_p^2=0.23$), in the forward direction EMG_{MEAN} was greater in stable ($p=0.006$) and unstable ($p=.001$) than unloaded, there was no difference between load conditions for right or left directions. There was an interaction effect for RF-BF CI ($F(2,26)=7.32$, $p<.001$, $\eta_p^2=0.36$) but there were no simple main effects. There were no load or direction effects for either CI pair and there was no difference between stable and unstable for any EMG variable during LOS trials.

[Table 3 here]

Discussion

This study has demonstrated that when carrying a stable or unstable load of 15% BM, postural $SWAY_{PL}$ and complexity are reduced during quiet standing and the LOS are reduced. However, no differences were found for postural control variables between stable and unstable during quiet standing. There was an increase in GM-TA coactivation in unstable compared to stable conditions and reduced RF-BF coactivation in unstable compared to unloaded conditions during quiet standing. Furthermore, load carriage increased RF activity during LOS.

The decrease in sway path length found in the present sample of older adults contrasted with previous findings in young adults where an increase in sway length, area and velocity are reported[1,5–7,12]. Furthermore, the decrease in postural sway complexity, as indicated by a reduced D_f, would suggest older adults adopt a more constrained strategy in response to the added inertia of the load. In contrast, Hur et al.[5] found in young adult firefighters the addition of load (5.4-9.1 kg) increased the randomness of postural sway, possibly as the participants, being healthy younger adults experienced in load carriage, did not require a constrained control strategy to compensate for the added load. Previous studies have demonstrated that postural sway complexity is reduced in older adults compared to young[27,28] and older fallers compared to non-fallers[29]. The findings of the present study therefore suggest that added load perturbs the neuromuscular system of older adults requiring altered control strategies which are associated with impaired postural control.

The reduced LOS_{REL} found in the present study is also indicative of a conservative postural control strategy adopted by older adults in loaded conditions. The findings of the present study contrast with those found for young adults carrying backpacks, where no alteration in LOS displacements were found compared to unloaded LOS[30]. However, in load carriage tasks with increased difficulty such as held above the head[31] or in a single hand[32] a reduction in the LOS is found. Together these findings suggest that when a load carriage task is sufficiently challenging the LOS are reduced to maintain balance. In older adults, a backpack load is sufficiently challenging to require a reduction in the LOS to maintain balance. Furthermore, smaller LOS values can retrospectively identify fallers and multiple fallers from non-fallers in older adult populations[33,34]. The results of this study therefore suggest that load carriage can increase the risk of falls in older adults as the distance the COM can be moved whilst maintaining stability is reduced. It could also be considered that the reduced LOS caused by load carriage in this study are the result of age related reduction in torque production capacity of the muscles about the ankle and/or hip joints. Reduced strength will also result in a more conservative postural control strategy, when loaded, to reduce the moment arm length of the COM and therefore the torque generated by gravity during the LOS task.

Contrary to the hypothesised effect, the present study found no difference between the stable and unstable load conditions for quiet standing or LOS variables. These findings are in contrast with previous findings where a handheld load that was unstable in the anterio-posterior direction increased COP displacement compared to a stable load in younger adults[3]. Since the unstable load used in the present study was comprised of water the perturbations generated by the load were small in magnitude. Participants were likely able to compensate for any instability. Interestingly, in unstable there was a greater GM-TA coactivation when compared stable possibly indicating that participants attempted to stiffen the ankle joint[18] in response to the unstable load.

The increase in GM and RF activity during quiet standing and LOS respectively, and reduction in RF-BF coactivation during quiet standing in loaded conditions compared to unloaded indicate that the demand on anti-gravity muscles is increased. However, these findings are in opposition to those of previous studies that reported no load carriage effects on lower limb muscle activation in younger adults[12,15]. It is possible that younger adults can accommodate the added load with changes in trunk muscle activity[15] without the needed for additional activity of the lower limbs. Furthermore, previous studies investigating the effect of load on muscle activations have not measured the activity of the Triceps Surae muscles[3,12,15]. The load effects on GM activation and GM-TA coactivation in the present study suggest these studies[3,12,15] may have missed important information regarding the neuromuscular contributions to postural control adaptations in loaded conditions. Finally, the increased activation of anti-gravity muscles in older adults in response to backpack loads could suggest that load carriage could be used as a physical training intervention to improve muscle strength in older adults. However, it is worth considering the acute impacts on postural control so this should be performed in controlled environments but may provide further beneficial adaptations to postural control when training regularly with loads.

There were limitations of the present study. Interpretation of the results are limited to community dwelling older adults and only to quiet standing conditions, however previous studies have investigated the effects during walking[21]. Future study should focus on the effects of load carriage on frail older adults and clinical populations as it may be expected that load carriage will have a greater effect on postural control in these populations which could have implications for fall risk. It may also be considered a limitation that the assessment of EMG activity was limited to muscles of the lower limb. It is likely that the trunk muscles play an important role in maintaining stability and producing neuromuscular compensation strategies under loaded conditions, particularly during LOS tests. It could also be considered that the Vastii muscles may also provide additional insight in the study of loaded postural control as key anti-gravity muscles. The effects of load carriage in older adults on these muscles should be considered an area of future research. In addition, the decision to normalise EMG signals to the maximum value in the unloaded condition can also affect the interpretation of coactivation values since the calculation of coactivation indices requires the assumption that the muscle with the largest activity is the agonist which may not be accurate when normalised. However, this approach does still allow for the comparison of overall coactivation between load conditions. Finally, since the average BMI of the included participants was 25.6 $kg\cdot m^{-2}$ the sample represents an overweight population, however only 1 participant would be considered obese with a BMI >30 $kg\cdot m^{-2}$. This should be taken into consideration when comparing the findings of the current study. However, given the within subjects design of the study and the use of a load relative to the BM of participants it is expected that the BMI of participants would limited effect on the present findings.

In conclusion, this study presents novel results demonstrating that when older adults carry a load equivalent to 15% BM postural sway magnitude and complexity during quiet standing are reduced. There was also a reduction in LOS which may indicate an increased risk of falls for older adults carrying loads. The results of the present study suggest that older adults adopt a constrained, conservative postural control strategy in loaded conditions. However, there was no difference in postural control between carrying a stable and unstable load. During quiet standing a greater GM activity was found in unstable than unloaded conditions and greater GM-TA coactivation in unstable than stable conditions, indicating greater anti-gravity muscle activity is required in loaded conditions and greater ankle stiffness is required in unstable load conditions. Furthermore, RF activity was greater when carrying a load during the LOS than unloaded.

Conflicts of interest

The authors declare there were no conflicts of interest.

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Figure Captions

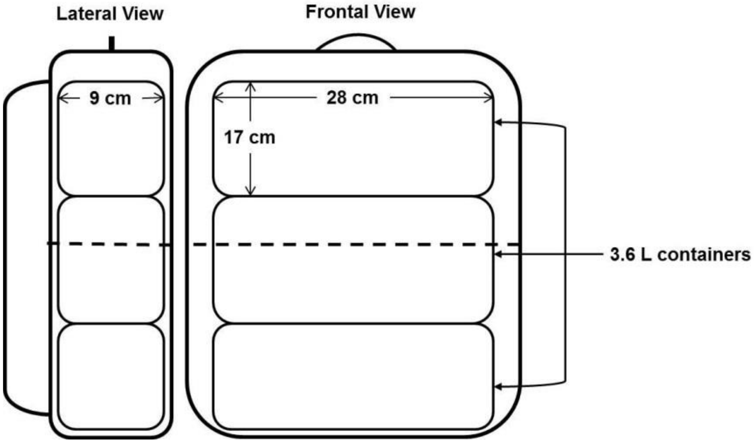

Figure 1. Illustration of the position of containers inside the backpack. Each container held either steel weights for the stable condition or steel weights and water for the unstable condition, distributed evenly between the 3 containers.

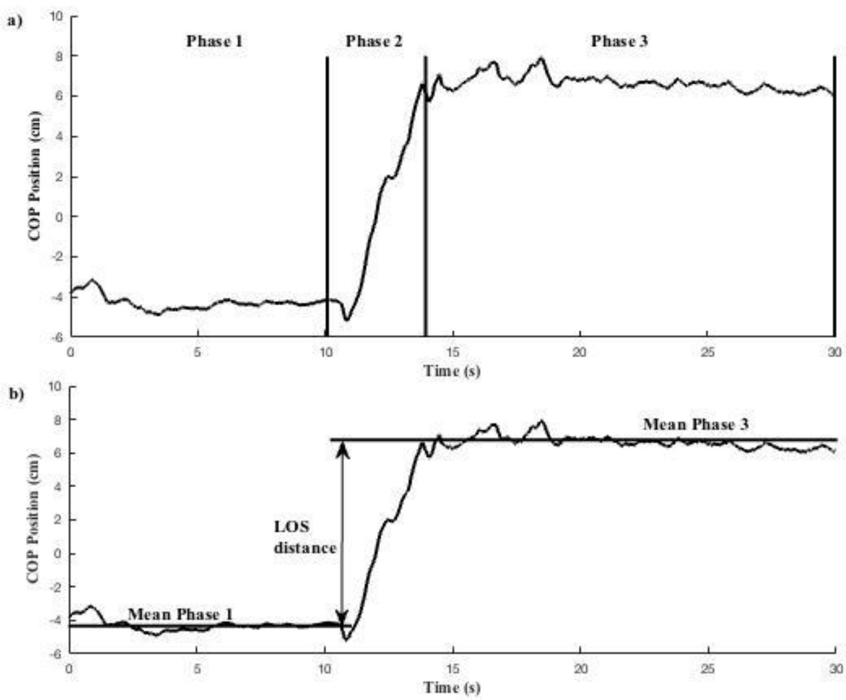

Figure 2. The a) phase definition of limits of stability (LOS) trials and b) LOS distance definition.

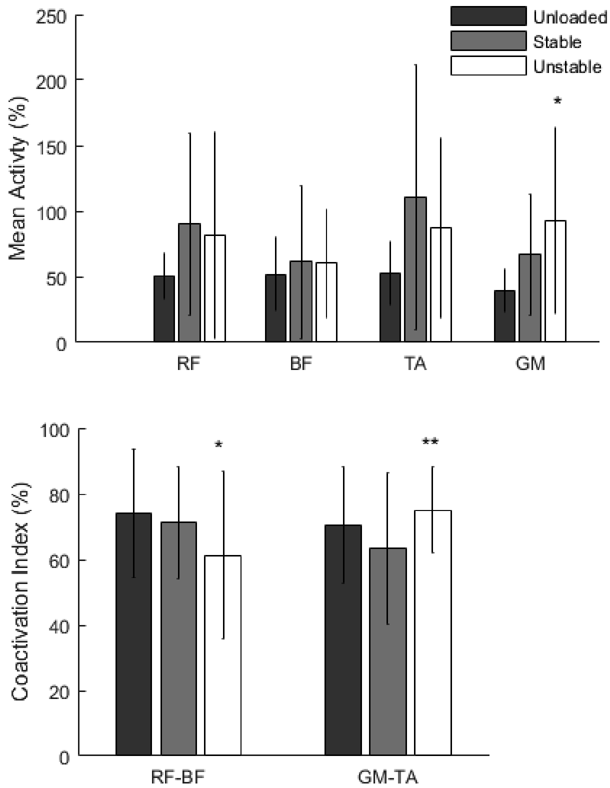

Figure 3. Mean and standard deviation values for the a) mean EMG activity and b) coactivation indices for all muscles and muscle pairs during quiet standing in the unloaded, stable and unstable load conditions.

* indicates the value is significantly different to unloaded condition, ** indicates value is significantly different to stable condition.

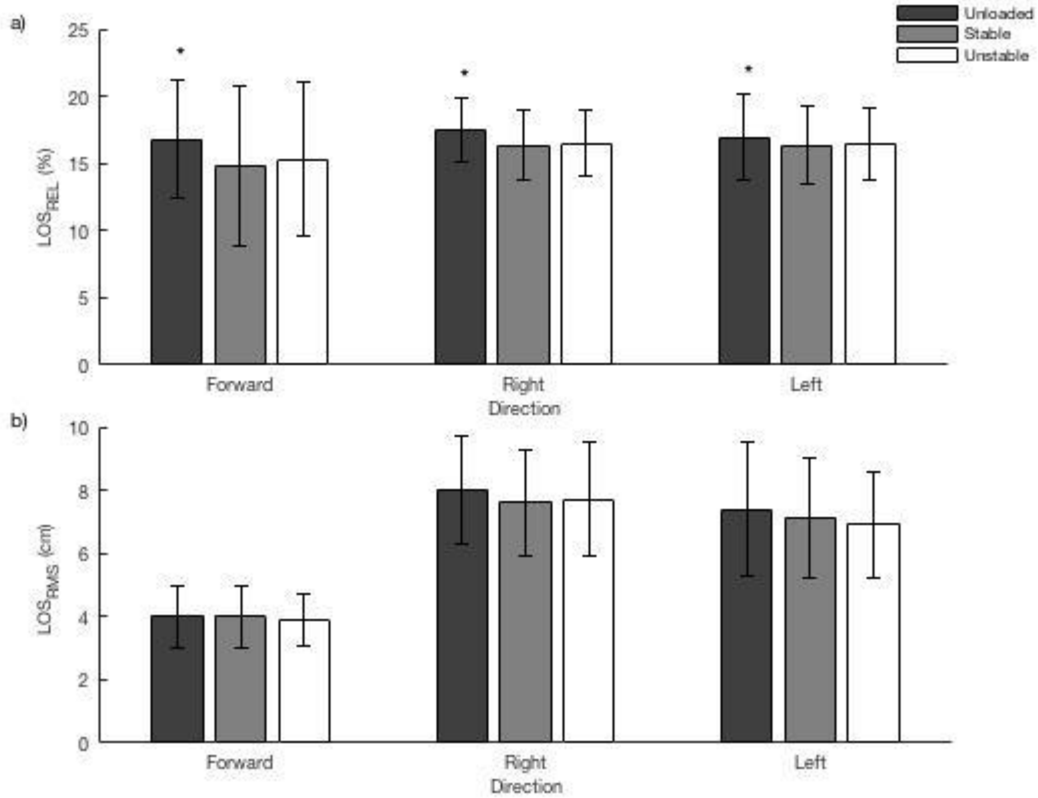

Figure 4. Mean and standard deviation values for a) limits of stability relative to base of support length (LOS_{REL}) and b) root mean square value during sustained leaning (LOS_{RMS}) for the forward, right and left directions in the unloaded, stable and unstable load conditions. * indicates that unloaded is greater than stable and unstable load conditions.

Tables

Table 1. Electrode placements for the 4 lower limb muscles studied.

Muscle	Electrode position
Rectus Femoris	50% along the line from the anterior superior iliac spine to the superior border of the patella.
Biceps Femoris	50% along the line between the ischial tuberosity and the lateral epicondyle of the tibia.
Tibialis Anterior	33% along the line between the tip of the fibula and the tip of the medial malleolus.
Gastrocnemius Medialis	Most prominent bulge of the muscle.

Table 2. Mean and standard deviation values for all quiet standing postural control variables in the unloaded, stable and unstable conditions.

Variable	Unloaded	Stable	Unstable
$SWAY_{PL}$ (cm)	94.5±18.9	81.3±15.8*	83.4±14.9*
ML D_f	1.8±0.1	1.6±0.1*	1.7±0.1*
AP D_f	1.5±0.1	1.4±0.1*	1.4±0.1*

* indicates the value is significantly different to unloaded condition

Table 3. Mean and standard deviation values for the mean EMG (EMG_{MEAN}) of all four muscles and coactivation index (CI) of both muscle pairs in each LOS direction in the unloaded, stable and unstable conditions.
* indicates stable and unstable were greater than unloaded, † indicates a significant interaction.

	Forward			Right			Left			
	Unloaded	Stable	Unstable	Unloaded	Stable	Unstable	Unloaded	Stable	Unstable	Effects
EMG_{MEAN} (%)										
RF	57.6±35.1	79.1±53.2	83.9±43.0	72.4±32.7	105.5±63.5	101.2±55.6	67.2±31.8	105.8±49.9	127.6±91.3	*
BF	126.2±57.2	126.7±73.3	102.3±50.2	97.1±56.6	105.9±67.8	109.7±58.1	87.6±41.9	89.6±43.4	70.5±56.8	
GM	69.5±31.5	122.9±49.0	107.5±51.2	56.0±31.1	76.2±32.3	77.0±46.9	99.9±36.3	116.2±44.2	93.2±76.3	
TA	70.3±36.2	118.4±74.2	135.5±74.9	188.7±98.2	166.6±62.9	169.6±86.5	114.2±43.6	151.7±60.1	190.6±138.6	†
CI (%)										
RF-BF	49.2±15.1	56.1±16.6	60.3±16.8	70.6±14.9	74.0±17.7	65.8±19.8	72.3±16.3	64.6±25.2	66.1±22.3	†
GM-TA	67.5±13.6	62.6±20.2	57.7±21.2	56.1±27.3	54.9±25.5	49.3±13.9	63.7±23.9	55.7±25.9	52.2±19.7	