

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

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24 **Abstract**

25 The aim of this study was to investigate the effects of backpack load carriage on quiet
26 standing postural control and limits of stability of older adults. Fourteen older adults (65±6
27 years) performed quiet standing and a forward, right and left limits of stability test in 3
28 conditions, unloaded, stable and unstable backpack loads while activity of 4 leg muscles was
29 recorded. Stable and unstable loads decreased postural sway (main effect $\eta_p^2=0.84$, stable:
30 $p<.001$, unstable: $p<.001$), medio-lateral (main effect $\eta_p^2=0.49$, stable: $p=.002$, unstable:
31 $p=.018$) and anterior-posterior (main effect $\eta_p^2=0.64$, stable: $p<.001$, unstable: $p=.001$) fractal
32 dimension and limits of stability distance (main effect $\eta_p^2=0.18$, stable: $p=.011$, unstable:
33 $p=.046$) compared to unloaded. Rectus Femoris (main effect $\eta_p^2=0.39$, stable: $p=.001$,
34 unstable: $p=.010$) and Gastrocnemius (main effect $\eta_p^2=0.30$, unstable: $p=.027$) activity
35 increased in loaded conditions during limits of stability and quiet standing. Gastrocnemius-
36 Tibialis Anterior coactivation was greater in unstable load than stable loaded quiet standing
37 (main effect $\eta_p^2=0.24$, $p=.040$). These findings suggest older adults adopt conservative
38 postural control strategies minimising the need for postural corrections in loaded conditions.
39 Reduced limits of stability may also increase fall risk when carrying a load. However, there
40 was no difference between unstable and stable loads for postural control variables.

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

44 **Word Count:** 3715

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Introduction

50 Disturbances to the postural control system can come from numerous sources including
51 physical perturbations, muscle fatigue and load carriage¹⁻³. It was demonstrated previously
52 that a period of prolonged walking can lead to postural control alterations in older adults⁴. A
53 potential explanation is that the fatigue results in acute changes to the force production
54 capabilities of a muscle resulting in a smaller muscle force production to body mass ratio.
55 This observation was supported by Ledin et al.,¹ although, they also found that load carriage
56 had a larger impact on postural control than muscle fatigue. Carrying a load on the trunk, e.g.
57 wearing a backpack, artificially increases the mass of the trunk. This negatively impacts the
58 ability to perform postural corrections as the force output needed for postural corrections is
59 increased⁵.

60 Previous studies investigating the effect of load carriage on postural control of
61 younger adults found increased postural sway^{1,5-7} and complexity of postural sways⁵. During
62 tasks requiring participants to move the centre of mass (COM) towards the limits of stability,
63 handheld loads reduce the maximum distance young adults can move the COM. Together
64 these findings suggest that load carriage reduces postural stability⁵ which could have
65 implications for fall risk in older adults.

66 Unstable loads have different effects on postural control than stable loads³, suggesting
67 the type of load can also impact postural control. An unstable load held in the hands increases
68 sway velocity and area in young adults³. In addition, older adults are more likely to be
69 affected by load carriage^{8,9}. Movements of an unstable load require individuals to produce
70 additional corrective forces to attenuate perturbation provided by the load, increasing the
71 demand on the postural control system. The use of perturbations to investigate the stability of
72 the postural control system is common^{10,11}. Load carriage perturbs the postural control system
73 by increasing the mass that must be supported and controlled^{1,5,12}, this effect can be

74 magnified by unstable loads³, providing insight into the mechanisms of postural control
75 adopted by older adults when perturbed. Non-linear measures of postural sway complexity,
76 such as the fractal dimension, can elucidate the neuromuscular control mechanisms^{13,14}
77 adopted when the system is perturbed.

78 Previous studies investigating the effect of load carriage on muscle activation in
79 young adults have focussed either on muscles of the trunk and upper leg^{3,12,15}. However,
80 these studies have not investigated the activation of Triceps Surae muscles which are largely
81 responsible for postural control^{16,17}. In addition, it has been suggested that older adults utilise
82 greater coactivation for postural control to compensate for age related neuromuscular
83 decline¹⁸. Older adults may therefore rely on increased coactivation in response to added
84 load.

85 It is currently unknown how load carriage affects older adults postural control, the limits
86 of stability and muscle activation. Load carriage is a common task for community dwelling
87 older adults and also provides a perturbation to the postural control system, therefore
88 allowing the study of the robustness of the postural control system to perturbations in a
89 commonly encountered paradigm¹⁹. The ability to respond to postural perturbations is
90 essential for minimising the risk of falls in older adults²⁰. To further explore the effect of
91 perturbations the current study included an unstable loaded condition. The aim of this study
92 was to determine the effect of stable and unstable load carriage on postural control, muscle
93 activation and coactivation during quiet standing and limits of stability tests in older adults. It
94 was hypothesised that stable and unstable load carriage would result in increased postural
95 sway magnitude and complexity, with concurrent increases in lower limb muscle activity and
96 coactivation. Additionally, it was hypothesised that stable and unstable loads would result in
97 decreased limits of stability length and increased variability, with a concurrent increase in the
98 lower limb muscle activity and coactivation. Finally, it was hypothesised that unstable loads

99 would have a greater effect on postural control, muscle activation and coactivation than
100 stable loads.

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Methods

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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·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.

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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 kg²¹. 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

123 15% of the participants BM. The order in which load conditions were performed was
124 randomised across participants.

125 [Figure 1 here]

126 Postural control during quiet standing and LOS were performed with participants
127 stood barefoot in a comfortable position on a force plate recording at 48 Hz (Kistler
128 Instruments Ltd, Winterthur, Switzerland) with eyes open. The foot position of each
129 participant was marked on a clear covering placed over the surface of the force plate to
130 ensure the same position was adopted for each trial, as foot placement can alter the calculated
131 postural sway parameters²².

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

143 [Figure 2 here]

144 During each quiet standing and LOS trial participants were fitted with reusable
145 bipolar electrodes with a 2 cm inter-electrode distance (SX230-1000, Biometrics Ltd, UK) to
146 measure the electromyographic (EMG) activity of the left Rectus Femoris (RF), Biceps
147 Femoris (BF), Tibialis Anterior (TA) and Gastrocnemius Medialis (GM). A reference

148 electrode was placed over the left radial head. Specific electrode placements are outlined in
 149 Table 1. The skin was prepared by shaving the area and cleaning with an alcohol wipe. The
 150 electrodes were attached to an 8-channel amplifier (range: $\pm 4\text{mV}$, gain: 1000, impedance:
 151 $1\text{M}\Omega$ - K800, Biometrics Ltd, UK) before being A/D converted (CA-1000, National
 152 Instruments Corp., UK).

153 [Table 1 here]

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

157 Quiet Standing: The recorded centre of pressure (COP) signals were not filtered to
 158 avoid removing the natural variability of the signal which would impact the non-linear
 159 analyses as the complexity of the signal is removed¹³. The postural sway path length
 160 (SWAY_{PL}) was calculated as the resultant path length of the medio-lateral (ML) and antero-
 161 posterior (AP) COP components. Fractal dimension (D_f) was calculated using Higuchi's
 162 algorithm²³ to estimate the complexity of the COP signals in the AP and ML directions. The
 163 time series $x=x(1),x(2),x(3),\dots,x(N)$ is reconstructed into k new time series, $x(m,k)$ with initial
 164 time value m , and discrete time interval k :

$$165 \quad x(m,k) = x(m), x(m+k), x(m+2k), \dots, x\left(m + \left\lfloor \frac{N-m}{k} \right\rfloor k\right)$$

$$166 \quad m = 1, 2, 3, \dots, k$$

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

$$171 \quad L_m(k) = \frac{\sum_{i=1}^{\lfloor (N-m)/k \rfloor} |x(m+ik) - x(m+(i-1)k)|}{\lfloor (N-m)/k \rfloor k}$$

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

$$176 \quad L(k) = \sum_{m=1}^k L_m(k)$$

177 The D_f is then determined as the slope of a linear least squares fit of the curve for $\ln(L(k))$ vs.
178 $\ln(1/k)$.

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

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

$$194 \quad LOS_{RMS} = \sqrt{\frac{1}{N} \sum_{n=1}^N |COP_n|^2}$$

195 where N is the length of the signal and COP_n is the n th element of the COP signal.

2196 Muscle Activation and Coactivation: Raw EMG signals were band-pass filtered with a
 2197 dual-pass 2nd order Butterworth filter with 20-450 Hz cut-off frequencies before being full-
 2198 wave rectified and low-pass filtered with a dual-pass 2nd order Butterworth filter with a 10
 2199 Hz cut-off frequency. Low-pass filtered EMG signals were normalised as a percentage of the
 2200 maximum activity recorded during 60 seconds of unloaded quiet standing²⁴. The average
 2201 activity of each muscle (EMG_{MEAN}) was calculated for each quiet standing and LOS trial
 2202 from the normalised signal.

2203 The coactivation indices²⁵ (CI) of 2 muscle pairs (RF-BF and GM-TA) were calculated as
 2204 follows:

$$2205 \quad CI = \frac{2I_{ant}}{I_{tot}} \times 100$$

2206 Where I_{tot} is the sum of the integrals of both muscles:

$$2207 \quad I_{tot} = \int_{t1}^{t2} [EMG_{agonist} + EMG_{antagonist}](t)dt$$

2208 and I_{ant} is the total integral of antagonistic activity, defined as the muscle with the lower
 2209 activity at each time point:

$$2210 \quad I_{ant} = \int_{t1}^{t2} EMG1(t)dt + \int_{t2}^{t3} EMG2(t)dt$$

2211 Where $t1$ and $t2$ denote periods that the activity of the first muscle of each pair is less than the
 2212 second, and $t2$ and $t3$ denote the periods that the activity of second muscle is less than the
 2213 first. Coactivation indices are expressed as a percentage of antagonistic activity with respect
 2214 to total activity for each pair.

2215 Statistics: All data were tested for normality using the Shapiro-Wilk test and for any
 2216 data that violated the assumption of sphericity the Greenhouse-Geiser correction was applied.
 2217 One-way repeated measures ANOVA were performed to determine the effect of load on quiet
 2218 standing postural control variables ($SWAY_{PL}$, $ML D_f$ and $AP D_f$) and muscle activation (RF,

219 BF, GM and TA, and RF-BF and GM-TA CI). To determine the effects of load condition,
220 direction and load x direction interaction effects on LOS variables (LOS_{REL} and LOS_{RMS}) and
221 muscle activation (RF, BF, GM and TA and RF-BF, and GM-TA CI) two-way repeated
222 measures ANOVA were performed. Post hoc pairwise comparisons with a Bonferroni
223 correction were performed for significant main effects. Simple main effects with Bonferroni
224 correction were used to explore significant interactions. For all tests $\alpha=0.05$ and partial eta
225 squared (η_p^2) was used as an estimate of effect size, values of 0.01, 0.06 and 0.14 were
226 interpreted as small medium and large effects respectively²⁶. All statistical analysis was
227 performed using SPSS software (v22, IBM UK Ltd., Portsmouth, UK).

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Results

230 There were significant load effects for SWAY_{PL} ($F(2,26)=68.75, p<.001, \eta_p^2=0.84$),
231 ML D_f ($F(2,26)=12.61, p<.001, \eta_p^2=0.49$) and AP D_f ($F(2,26)=23.13, p<.001, \eta_p^2=0.64$). All
232 quiet standing variables were greater in unloaded compared to stable (SWAY_{PL}: $p<.001$, ML
233 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:
234 $p=.001$) conditions. There were no differences between stable and unstable conditions (Table
235 2).

236 [Table 2 here]

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

243 [Figure 3 here]

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

248 [Figure 4 here]

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

258 [Table 3 here]

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Discussion

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

268 The decrease in sway path length found in the present sample of older adults contrasted

269 with previous findings in young adults where an increase in sway length, area and velocity
270 are reported^{1,5-7,12}. Furthermore, the decrease in postural sway complexity, as indicated by a
271 reduced D_f , would suggest older adults adopt a more constrained strategy in response to the
272 added inertia of the load. In contrast, Hur et al.⁵ found in young adult firefighters the addition
273 of load (5.4-9.1 kg) increased the randomness of postural sway, possibly as the participants,
274 being healthy younger adults experienced in load carriage, did not require a constrained
275 control strategy to compensate for the added load. Previous studies have demonstrated that
276 postural sway complexity is reduced in older adults compared to young^{27,28} and older fallers
277 compared to non-fallers²⁹. The findings of the present study therefore suggest that added load
278 perturbs the neuromuscular system of older adults requiring altered control strategies which
279 are associated with impaired postural control.

280 The reduced LOS_{REL} found in the present study is also indicative of a conservative
281 postural control strategy adopted by older adults in loaded conditions. The findings of the
282 present study contrast with those found for young adults carrying backpacks, where no
283 alteration in LOS displacements were found compared to unloaded LOS³⁰. However, in load
284 carriage tasks with increased difficulty such as held above the head³¹ or in a single hand³² a
285 reduction in the LOS is found. Together these findings suggest that when a load carriage task
286 is sufficiently challenging the LOS are reduced to maintain balance. In older adults, a
287 backpack load is sufficiently challenging to require a reduction in the LOS to maintain
288 balance. Furthermore, smaller LOS values can retrospectively identify fallers and multiple
289 fallers from non-fallers in older adult populations^{33,34}. The results of this study therefore
290 suggest that load carriage can increase the risk of falls in older adults as the distance the
291 COM can be moved whilst maintaining stability is reduced. It could also be considered that
292 the reduced LOS caused by load carriage in this study are the result of age related reduction
293 in torque production capacity of the muscles about the ankle and/or hip joints. Reduced

294 strength will also result in a more conservative postural control strategy, when loaded, to
295 reduce the moment arm length of the COM and therefore the torque generated by gravity
296 during the LOS task.

297 Contrary to the hypothesised effect, the present study found no difference between the
298 stable and unstable load conditions for quiet standing or LOS variables. These findings are in
299 contrast with previous findings where a handheld load that was unstable in the antero-
300 posterior direction increased COP displacement compared to a stable load in younger adults³.
301 Since the unstable load used in the present study was comprised of water the perturbations
302 generated by the load were small in magnitude. Participants were likely able to compensate
303 for any instability. Interestingly, in unstable there was a greater GM-TA coactivation when
304 compared stable possibly indicating that participants attempted to stiffen the ankle joint¹⁸ in
305 response to the unstable load.

306 The increase in GM and RF activity during quiet standing and LOS respectively, and
307 reduction in RF-BF coactivation during quiet standing in loaded conditions compared to
308 unloaded indicate that the demand on anti-gravity muscles is increased. However, these
309 findings are in opposition to those of previous studies that reported no load carriage effects
310 on lower limb muscle activation in younger adults^{12,15}. It is possible that younger adults can
311 accommodate the added load with changes in trunk muscle activity¹⁵ without the needed for
312 additional activity of the lower limbs. Furthermore, previous studies investigating the effect
313 of load on muscle activations have not measured the activity of the Triceps Surae
314 muscles^{3,12,15}. The load effects on GM activation and GM-TA coactivation in the present
315 study suggest these studies^{3,12,15} may have missed important information regarding the
316 neuromuscular contributions to postural control adaptations in loaded conditions. Finally, the
317 increased activation of anti-gravity muscles in older adults in response to backpack loads
318 could suggest that load carriage could be used as a physical training intervention to improve

319 muscle strength in older adults. However, it is worth considering the acute impacts on
320 postural control so this should be performed in controlled environments but may provide
321 further beneficial adaptations to postural control when training regularly with loads.

322 There were limitations of the present study. Interpretation of the results are limited to
323 community dwelling older adults and only to quiet standing conditions, however previous
324 studies have investigated the effects during walking²¹. Future study should focus on the
325 effects of load carriage on frail older adults and clinical populations as it may be expected
326 that load carriage will have a greater effect on postural control in these populations which
327 could have implications for fall risk. It may also be considered a limitation that the
328 assessment of EMG activity was limited to muscles of the lower limb. It is likely that the
329 trunk muscles play an important role in maintaining stability and producing neuromuscular
330 compensation strategies under loaded conditions, particularly during LOS tests. It could also
331 be considered that the Vastii muscles may also provide additional insight in the study of
332 loaded postural control as key anti-gravity muscles. The effects of load carriage in older
333 adults on these muscles should be considered an area of future research. In addition, the
334 decision to normalise EMG signals to the maximum value in the unloaded condition can also
335 affect the interpretation of coactivation values since the calculation of coactivation indices
336 requires the assumption that the muscle with the largest activity is the agonist which may not
337 be accurate when normalised. However, this approach does still allow for the comparison of
338 overall coactivation between load conditions. Finally, since the average BMI of the included
339 participants was 25.6 kg·m⁻² the sample represents an overweight population, however only 1
340 participant would be considered obese with a BMI >30 kg·m⁻². This should be taken into
341 consideration when comparing the findings of the current study. However, given the within
342 subjects design of the study and the use of a load relative to the BM of participants it is
343 expected that the BMI of participants would limited effect on the present findings.

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

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356 **Conflicts of interest**

357 The authors declare there were no conflicts of interest.

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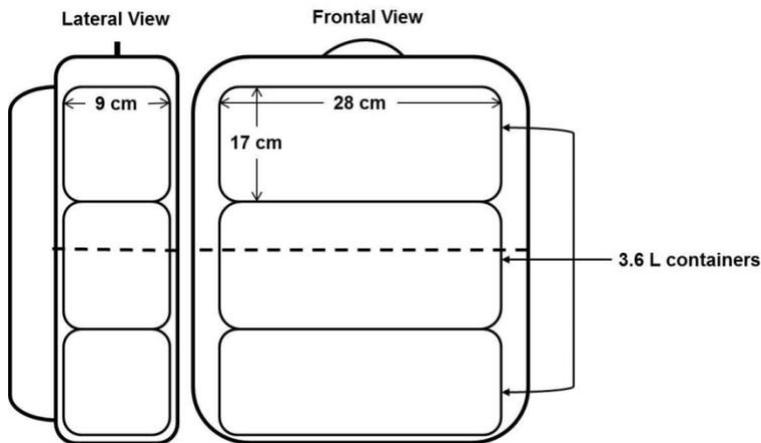
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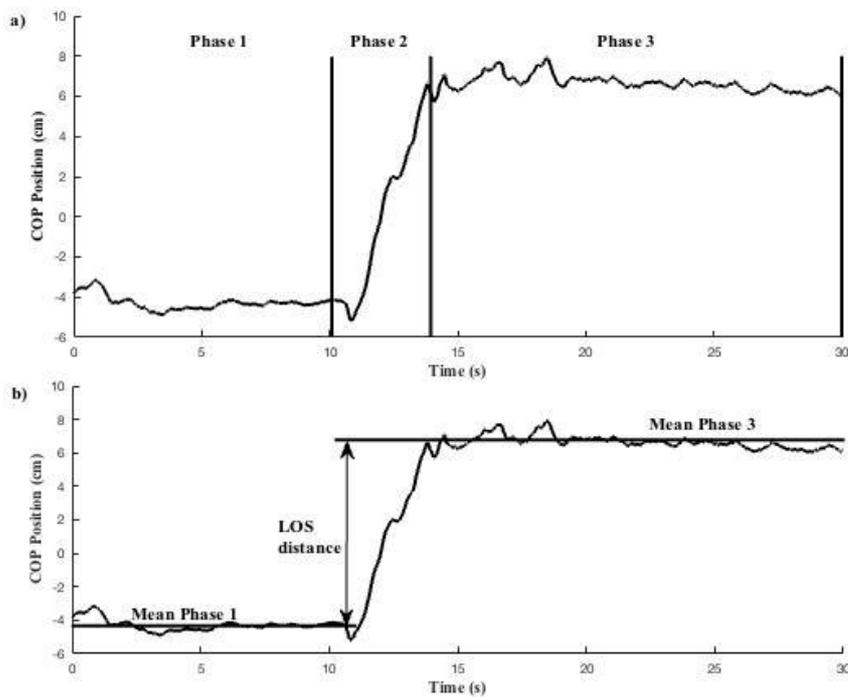
458 **Figure Captions**



459

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

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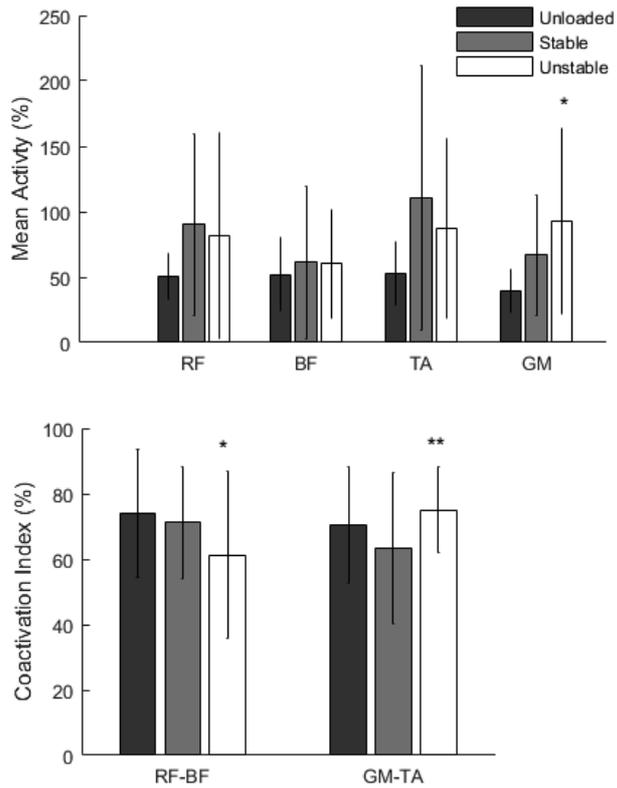


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465 Figure 2. The a) phase definition of limits of stability (LOS) trials and b) LOS distance
466 definition.

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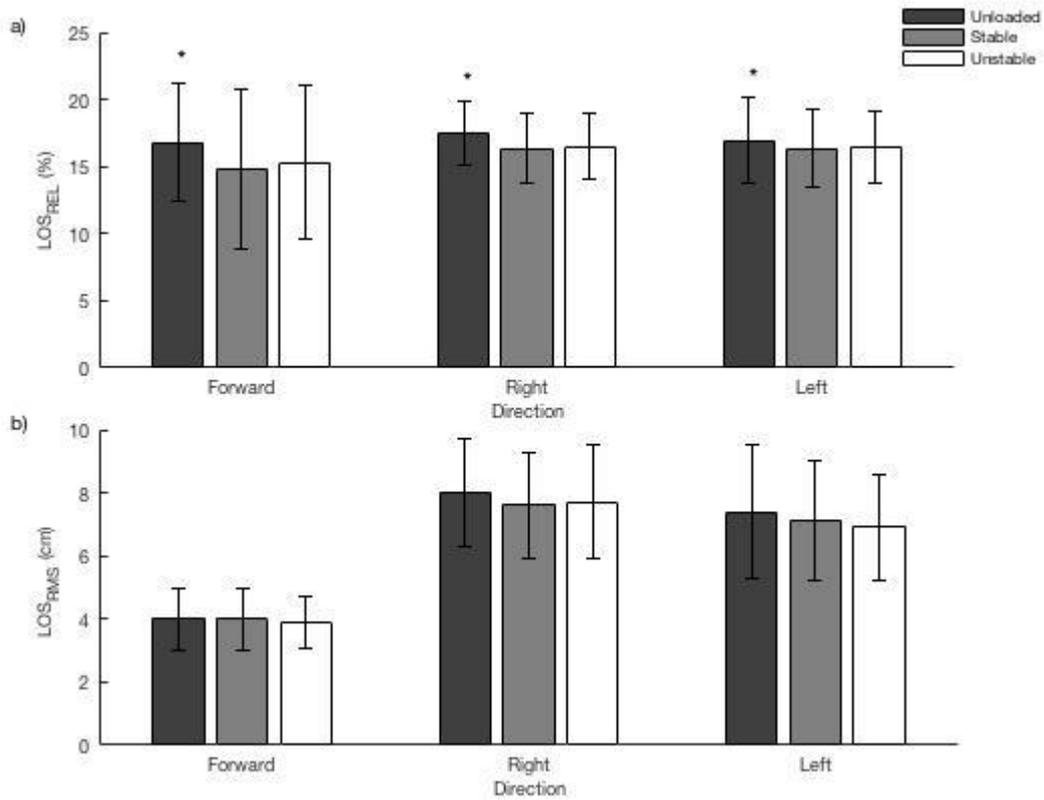


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470 Figure 3. Mean and standard deviation values for the a) mean EMG activity and b)
 471 coactivation indices for all muscles and muscle pairs during quiet standing in the unloaded,
 472 stable and unstable load conditions.

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

475



476

477 Figure 4. Mean and standard deviation values for a) limits of stability relative to base of
 478 support length (LOS_{REL}) and b) root mean square value during sustained leaning (LOS_{RMS})

479 for the forward, right and left directions in the unloaded, stable and unstable load conditions.

480 * indicates that unloaded is greater than stable and unstable load conditions.

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482

483 **Tables**

484 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.

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486

487 Table 2. Mean and standard deviation values for all quiet standing postural control variables
488 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*

489 * indicates the value is significantly different to unloaded condition

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501 Table 3. Mean and standard deviation values for the mean EMG (EMG_{MEAN}) of all four
 502 muscles and coactivation index (CI) of both muscle pairs in each LOS direction in the
 503 unloaded, stable and unstable conditions.

504 * indicates stable and unstable were greater than unloaded, † indicates a significant
 505 interaction.

	Forward			Right			Left			Effects
	Unload ed	Stable	Unstabl e	Unload ed	Stable	Unstabl e	Unload ed	Stable	Unstabl e	
EMG_{MEAN} (%)										
RF	57.6±3 5.1	79.1±5 3.2	83.9±4 3.0	72.4±3 2.7	105.5± 63.5	101.2± 55.6	67.2±3 1.8	105.8± 49.9	127.6± 91.3	* †
BF	126.2± 57.2	126.7± 73.3	102.3± 50.2	97.1±5 6.6	105.9± 67.8	109.7± 58.1	87.6±4 1.9	89.6±4 3.4	70.5±5 6.8	
GM	69.5±3 1.5	122.9± 49.0	107.5± 51.2	56.0±3 1.1	76.2±3 2.3	77.0±4 6.9	99.9±3 6.3	116.2± 44.2	93.2±7 6.3	
TA	70.3±3 6.2	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±1 5.1	56.1±1 6.6	60.3±1 6.8	70.6±1 4.9	74.0±1 7.7	65.8±1 9.8	72.3±1 6.3	64.6±2 5.2	66.1±2 2.3	†
GM-TA	67.5±1 3.6	62.6±2 0.2	57.7±2 1.2	56.1±2 7.3	54.9±2 5.5	49.3±1 3.9	63.7±2 3.9	55.7±2 5.9	52.2±1 9.7	