Changes in postural sway and gait characteristics as a consequence of anterior load carriage

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ABSTRACT

Background Anterior load carriage represents a common daily and occupational activity. Carrying loads in front of the body could potentially increase instability during standing and walking. **Research question** This study examined the effects of anterior load carriage on postural sway and gait parameters in healthy adults. **Methods** Twenty-nine participants (19 males, 10 females, age = 33.8 ± 12.7 years, height = 1.73 ± 0.07 m, mass = 75.1 ± 13.7 kg) were assessed in four conditions; (1) carrying no load (CON), (2) carrying a load with no added weight (i.e. empty box), (3) carrying a load with 5% body mass, and (4) carrying a load with 10% body mass. Anteroposterior and mediolateral centre of pressure (COP) displacement (cm) and the mean COP velocity ($cm \cdot s^{-1}$) were used to characterise postural sway. Coefficient of variation of the stride length, stride time and double support time were calculated from 1 min of treadmill walking at a preferred pace for gait assessment. **Results** The addition of the 10% load elicited an increase in anteroposterior COP displacement when compared to CON (d = 1.59), 0% (d = 1.50), and 5% (d = 0.75) ($P \leq 0.001$). The addition of the 10% load increased stride time (d = 1.71) and stride length (d = 1.20) variability when compared to CON ($P \leq 0.001$). **Significance** In summary, the increase in postural sway and gait variability with added weight is dependent on the magnitude of the load, where the greater the load, the greater the effect on static and dynamic stability. Anterior load carriage potentially increases the risk of fall-related injuries.

Key words: Anterior external loads · Functional task · Walking · Balance ·

HIGHLIGHTS

- We assessed postural sway and gait variability during anterior load carriage
- Postural sway and gait variability increased when holding an anterior external load
- The increase in postural sway and gait variability is dependent on the load magnitude
- Individuals should take caution when carrying loads greater than 5% body mass

1. INTRODUCTION

Previous research has indicated that carrying externals loads elicits an increase in centre of pressure (COP) measures of postural sway during quiet standing [1-5] and stride-to-stride gait variability during walking [6,7]. These changes are important as load carriage during standing and walking is a common practice in different occupational and daily tasks [8]. Therefore, further investigation of how postural stability is influenced by external loads is warranted.

To date, the vast majority of research that has investigated the influence of external loading on postural sway and gait parameters has examined posterior (e.g. backpack) [1,4,5,8] or lateral (e.g., grocery bags) load carriage [2,9-11]. Carrying external loads changes the mass-inertia characteristics of the body's centre of mass (COM). For example, when a backpack is added, the combined COM of the backpack and body shifts posteriorly, which is compensated for by a forward trunk lean to maintain the position of the body and load COM over the base of support [11]. Posterior load placement has been shown to elicit an increase in anteroposterior postural sway [1,4,5,8]. Additionally, carrying a backpack induces a slower walking velocity and increases double support time [12,13] and gait variability [7], reflecting reduced gait stability [14]. Accordingly, closer examination of balance and gait characteristics during different load carriage scenarios is essential.

Despite these initial enquires into the effects of carrying external loads on postural stability, little research has examined the effects of anterior load carriage (i.e., carrying loads with the hands and forearms). This gap in the literature is important as many daily (e.g. carrying a laundry basket) and occupational (e.g. courier delivery) activities require loads to be carried in front of the body. To further develop our understanding of the effects of external loads on postural stability, investigation is necessary to determine how anterior loads modulate changes in postural sway and gait characteristics. Only one study has reported the effects of anterior load carriage on postural sway [15]. It was shown that holding a box with 10% body mass in front of the body increased postural sway [15]. However, Shigaki et al. [15] examined only one load (i.e. 10% body mass). Given that impairments in postural sway [4] and gait stability [7] are proportional to the mass of the backpack, it seems well justified to investigate increasing anterior load mass on these stability metrics.

Studies examining the interaction between anterior load carriage and gait are scarce, however a study by Perry *et al.* [16] assessed the effect of anterior load carriage on obstacle-crossing behaviour. The authors reported that participants increased obstacle toe clearance when carrying an anterior load, which may have been influenced by availability of visual information regarding obstacle position. In addition, load carried with both hands removes the option of arm swing to counteract unbalanced loads on the body [17], which is an important restriction as arm swing contributes to overall gait stability [18]. Thus, there is a reasonable theoretical basis for expectation that anterior load carriage involving both arms will impair gait stability.

Within this context, to further develop our understanding of anterior load carriage, this proof of concept study examined the effects of increasing loads carried anteriorly on postural sway and gait parameters in healthy adults. We hypothesised that increases in COP measures of postural sway would be proportional to the load added to a box. Additionally, we also hypothesised that with increasing load, greater gait instability would be observed, characterised by changes in mean gait metrics and an increased gait variability. Understanding changes in postural sway and gait stability with anterior load carriage may contribute to the development of occupational training interventions aimed to mitigate the potentially negative effects of this type of loading on fall-risk. The present findings may also be influential in guiding future efforts to improve ergonomic design, such as on-body assistive devices to reduce both metabolic stress and balance impairments.

2. METHODS

2.1 Participants

Twenty-nine healthy participants (19 males, 10 females age; 33.8 ± 12.7 years [18 - 54 years], height; 1.73 ± 0.07 m, mass; 75.1 ± 13.7 kg) volunteered to participate in this study after providing written, informed consent. Exclusion criteria included: age $\geq$ 60 years, a history of lower back pain or lower back injury, and any neurological, musculoskeletal, orthopaedic and/or cardiovascular or pulmonary diseases that might affect balance or gait. The study was carried out in accordance with the guidelines outlined in the declaration of Helsinki (1964) and the study procedures were approved by the institutional ethics committee.

2.2 Procedures

A semi-randomised controlled cross-over study design was employed with each participant visiting the laboratory on two separate occasions in a counterbalanced order; (1) static postural sway assessment and, (2) gait assessment. Participants performed four conditions during each assessment; (1) control with no load (CON), (2) anterior load with no added weight (0%, empty box weight; 1.5 kg), (3) anterior load with 5% body mass (3.75 ± 0.68 kg) and (4) anterior load with 10% body mass (7.51 ± 1.37 kg). The within session order of task conditions were randomly assigned using Research Randomizer (www.randomizer.org). Each box (external dimensions; L 48cm × W 39 cm × D 20 cm, internal dimensions; L 39.5cm × W 33.5 cm × D 17 cm, volume; 22.5 L, mass; 1.5 kg) was filled with sealed bags of sand to ensure the distribution of mass was relatively uniform and to prevent excessive movement of the load when walking. During loaded and unloaded conditions, participants were instructed to fix their eyes ahead at a point on the wall and to hold the box against their abdomen with elbows flexed at 90º [16].

2.3 Static postural stability assessment

To examine the effects of load magnitude on postural sway each participant performed quiet stance trials while standing on a force platform (AMTI, AccuGait, Watertown, MA) for 30 s. Data were sampled at 100 Hz (AMTI, Netforce, Watertown, MA) and the maximal displacement of the COP in the anteroposterior and mediolateral directions (cm) and mean COP velocity ($cm \cdot s^{-1}$) were subsequently calculated (AMTI, BioAnalysis, Version 2.2, Watertown, MA). Participants were asked to stand as still as possible on the force platform with their feet together, arms by their sides (CON), while gazing at a target 1.5 meters from the force platform. Participants practiced each postural task once prior to recorded trials. A total of three trials were recorded consecutively for each condition and the mean of these trials was used in subsequent analysis.

2.4 Gait assessment

Gait stability determined during steady-state walking on a treadmill (h/p/Cosmos, Gaitway, Traunstein, Germany) using two in-dwelling force platforms (Kistler, Winterthur, Switzerland) (Fig. 1). Participants walked with their own footwear at a self-selected speed (4.21 ± 0.26 km/h). We asked participants to wear comfortable walking shoes, but not shoes with a heel. To habituate participants to walking on the treadmill and to ascertain self-selected walking speed for subsequent trials, each participant walked for 5-10 min on the treadmill. Participants were specifically instructed to walk at a preferred comfortable pace. All participants were blind to their self-selected walking speed, and the principal investigator adjusted speed in 0.2 km/h increments in response to instructions from the participant to go “slower” or “faster”. The principal investigator stood next to the treadmill to assist the participants to complete the tests safely. Ground reaction forces were sampled at 200 Hz, enabling the acquisition of stride time (sec), stride length (m) and double-limb support time (sec). The coefficient of variation (CV; [SD/Mean]*100) was also calculated for each of the gait indices to assess gait variability, a marker of gait instability and fall-risk [19]. For each load condition, participants walked on the treadmill for 2 min, with the final minute recorded for analysis. An average of all strides in the 1 min period were used in subsequent analyses.

Heart rate (HR) was continually monitored (Polar Electro, Oy, Finland) and recorded in the final 10 s of each loaded gait condition. A rating of perceived exertion (RPE) for both the working muscles (RPE_L) and central cardiorespiratory stress (RPE_C) using the 6 – 20 point Borg scale [20] was obtained at the same time as HR.

***** FIGURE 1 ABOUT HERE *****

2.6 Statistical analyses

An *a priori* power analysis (variable = mean COP velocity, power = 0.80, alpha = 0.05, effect size = 1.2) calculated from similar research (i.e. [2]) revealed a minimum of 10 participants was sufficient for finding statistically significant effects of load magnitude. For all analyses, normality (Shapiro–Wilk test) and homogeneity of variance/sphericity (Levene's test) were performed and confirmed prior to parametric tests. Separate repeated measures analysis of variance (ANOVA) were used to examine the differences in COP measures of postural sway and gait characteristics among the different load conditions. Where significant differences were detected, post hoc analyses using Bonferroni correction determined the location of any differences. Cohen's *d* effect sizes are reported for post hoc comparisons. Statistical significance was accepted at $P \leq 0.05$ for all tests. Statistical analyses were carried out using SPSS version 24.0 software (IBM Inc., Chicago, IL).

3. RESULTS

3.1 Static postural stability assessment

There was a main effect of load magnitude on anteroposterior COP displacement ($F_{(3,84)}$ = 104.908, $P \leq 0.001$) and mean COP velocity ($F_{(3,84)}$ = 25.727, $P \leq 0.001$) (Fig. 2). Post hoc analyses revealed that when compared to CON, anteroposterior COP displacement increased with the addition of a 5% ($P \leq 0.001$, $d = 0.74$) and 10% ($P \leq 0.001$, $d = 1.59$) load. The anteroposterior COP displacement was also greater during the 5% ($P \leq 0.001$, $d = 0.64$) and 10% ($P \leq 0.001$, $d = 1.50$), when compared to the 0% load. Finally, anteroposterior COP displacement increased from the 5% to the 10% load ($P \leq 0.001$, $d = 0.75$). The mean COP velocity was greater during the 10% load compared to CON ($P \leq 0.001$, $d = 0.90$), 0% ($P \leq 0.001$, $d = 0.80$) and 5% ($P \leq 0.001$, $d = 0.75$). No load effects were reported for mediolateral COP displacement ($F_{(3,84)}$ = 1.258, $P = 0.294$).

***** FIGURE 2 ABOUT HERE *****

3.2 Gait assessment

There was a main effect of load magnitude on stride time variability ($F_{(3,84)}$ = 56.115, $P \leq 0.001$) and stride length variability ($F_{(3,84)}$ = 15.486, $P \leq 0.001$) (Fig. 3). Post hoc analyses showed that when compared to CON, stride time variability was increased during the 5% ($P \leq 0.001$, $d = 0.89$) and 10% ($P \leq 0.001$, $d = 1.71$) trials. Stride-time variability was also greater during 10% compared to 0% ($P \leq 0.001$, $d = 1.54$). Stride length variability was greater during 10% compared to CON ($P = 0.001$, $d = 1.20$), 0% ($P = 0.004$, $d = 1.08$) and 5% ($P = 0.001$, $d = 1.01$). No load effects were reported for double support time variability ($F_{(3,84)}$ = 1.333, $P = 0.269$). No load effects were observed for mean stride length ($F_{(3,112)}$ =

.766, $P = 0.515$), stride time ($F_{(3,112)} = 1.228$, $P = 0.303$) or double support time ($F_{(3,112)} = .025$, $P = 0.995$) (Fig. 4).

***** FIGURE 3 ABOUT HERE *****

***** FIGURE 4 ABOUT HERE *****

3.3 Physiological responses

There was a main effect of load mass on HR ($F_{(3,84)} = 30.586$, $P \leq 0.001$), RPE_C ($F_{(3,84)} = 39.818$, $P \leq 0.001$) and RPE_L ($F_{(3,84)} = 47.957$, $P \leq 0.001$). Heart rate was greater during the 10% load compared to CON ($d = 1.04$), 0% ($d = 0.68$) and 5% ($d = 0.54$) conditions ($P \leq 0.001$) (Table 1). Heart rate was also greater during 5% compared to CON ($P \leq 0.001$, $d = 0.53$). RPE_c was greater during the 10% load compared to CON ($d = 1.50$), 0% and 5% (both $d = 1.0$) loads ($P \leq 0.05$). RPE_C was also greater during 0% and 5% (both $d = 0.50$) when compared to CON ($P \leq 0.05$). Local RPE (arms and legs) was greater during the 10% load compared to CON ($d = 2.53$), 0% ($d = 2.53$) and 5% ($d = 1.50$) loads ($P \leq 0.001$). RPE_L was also greater 5% ($d = 1.53$) when compared to CON ($P = 0.012$).

***** TABLE 1 ABOUT HERE *****

4. DISCUSSION

The current findings indicate; (1) the anteroposterior COP displacement increased proportionally with greater anterior load mass (2) carrying heavy loads (10% body mass) significantly increased stride-to-stride variability (stride length and stride time), but not mean gait parameters, partly confirming our initial hypothesis. These findings have important implications because anterior load carriage is a common challenge during many occupational and daily activities.

4.1 Static postural stability assessment

The current data set extends the anterior load literature [15], by demonstrating that increases in postural sway are proportionally dependent on the load mass. A change in the mass-inertia characteristics of the body is one of several mechanisms that have been offered to explain the increased postural sway during quiet standing when holding a backpack [1,4,5,8]. During quiet stance without external loading, the line of gravity acts through the COM and centroid of the base of support, creating a relatively stable system [1]. However, the horizontal position of the COM can be expected to differ depending on the position of the load on the body. When carrying a backpack there is a posterior shift in the position the body plus load COM relative to the base of support, which is compensated for by forward trunk lean to move the body and load COM anteriorly [21]. Such a shift in the body plus load COM implies that holding a load in front of the body elicits an anterior shift in the horizontal position of the body plus load COM, which is likely compensated for by a backwards trunk lean to move the body plus load COM posteriorly. It is also important to note that the mass of the hands and forearms (~2-3% of total body mass) were also shifted forward from the midline of the body to hold the load. However, the 0% load condition (i.e., empty box) did not influence postural sway or gait stability, indicating that the relatively light mass of the arms and the empty box were insufficient to elicit instability in standing and walking.

From a physiological perspective, holding an external load intensifies cardiac and respiratory muscular contractions, which can theoretically increase postural sway [22]. Additionally, in the present study participants were instructed to hold the box against their abdomen, which may alter breathing mechanics by restricting the anterior regions of the thorax by imposing a volume limitation on the chest wall.

4.2 Gait assessment

Another main finding of the present study was that anterior load carriage increased stride-to-stride variability but not mean gait metrics, partly confirming our hypothesis. This is an important finding because most falls occur during dynamic activities [23] and fallers tend to demonstrate greater gait variability compared to non-fallers [24]. The present results concerning the moderate to large increases in stride length and time variability when carrying an anterior external load are not consistent with studies using posterior load carriage. Qu and Yeo [7] for example, reported that carrying a backpack load increased stride width variability, but not stride length variability. Stride length and time variability are thought to reflect gait timing mechanisms and pattern generator of gait, whereas stride width and double support time more closely reflect balance control mechanisms [25]. Thus, the increases in stride length and stride time variability seen here, may be related to changes in the rhythmicity control of walking but not balance function during gait. This discrepancy between studies may be due to two unique factors associated with anterior load carriage. Firstly, previous research has demonstrated that visual information from the lower visual field serves to modify lower limb trajectory and foot placement during gait [26]. However, in the present study, the lower visual field was obscured by the load. Further, one method of counteracting unbalanced loads on the body is to swing the arms [17]. In the present study arm movements were restricted by carrying the load, potentially contributing to the increased gait variability seen here.

Notably, changes in gait with external loads were observed for variability metrics (stride length and time), but not for the mean parameters of stride length, stride time or double support time. It has been reported that variability metrics are regulated independently of mean values [27]. Given that the regulation of variability is normally automated and requires minimal cognitive input [28], it is reasonable to hypothesise that carrying loads in front of the body requires at least some attentional-resources. The changes in variability but not mean gait metrics in the present study with external loads also likely reflect the greater sensitivity of variability parameters.

From a physiological perspective, load carriage is a physically demanding task. Although it has been reported that energy expenditure during walking increases linearly with an increase of anterior load magnitude [29], the physical demand of the anterior load carriage was well below individual's maximal physical capacity in the present study (based on RPE and % HR_{MAX}). Thus, changes in body dynamics (e.g. behaviour and location of the COM) are likely the primary factor contributing to reduced gait stability.

4.3 Limitations

The inclusion of only young and intermediate age groups precludes us from generalising these findings to older populations. We also only tested loads up to 10% body mass, which precludes us from making generalisations to heavier loads. Finally, gait stability was measured on a level ground motorised treadmill, which is dissimilar to 'real world' terrain (e.g., uneven surfaces, obstacles, steps and changes of direction). These factors limit the ecological validity of the gait outcomes and further study of complimentary measures (i.e. muscle activity and joint kinematics) are recommended.

5. CONCLUSION

Given that increased postural sway [30] and gait variability [19] are indicators of greater fall-risk, carrying heavier loads in front of the body may increase the risk of falls during static and dynamic situations. It is recommended that individuals take caution when carrying loads greater than 5% body mass in front of the body. It is anticipated that older adults will be at a significantly greater risk compared to the young and intermediate age groups due to increased gait variability and impaired static balance; future research should examine this issue in an ageing population.

Conflict of interest statement

None

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Figure 1: Participant walking on the treadmill while carrying anterior load

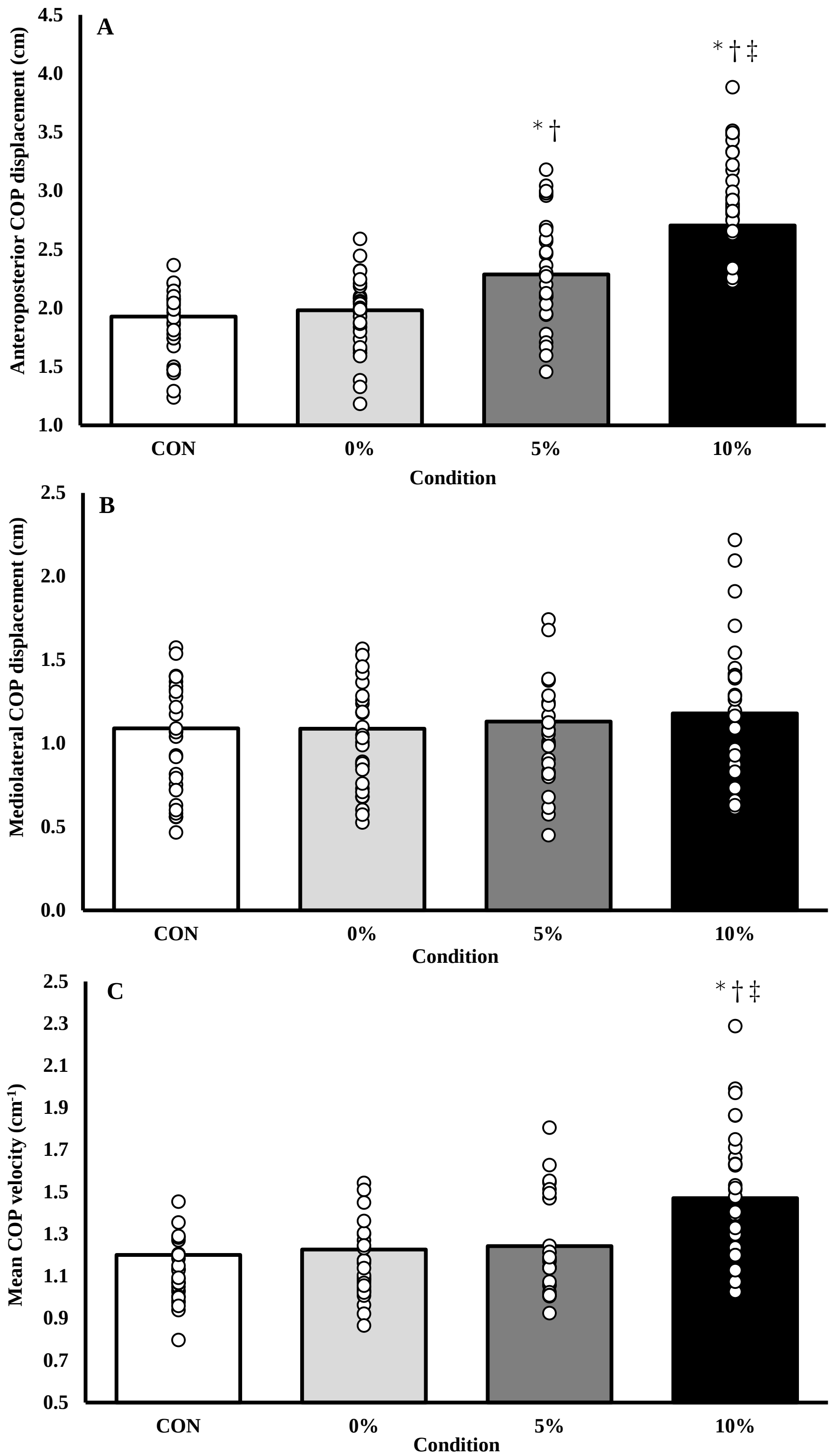

Fig. 2 Mean ± SD anteroposterior (A) and mediolateral (B) COP displacement, and COP velocity (C) during quiet standing. *Significantly different to CON. †Significantly different to 0%. ‡Significantly different to 5%. (P < 0.05). ○ Represent individual data.

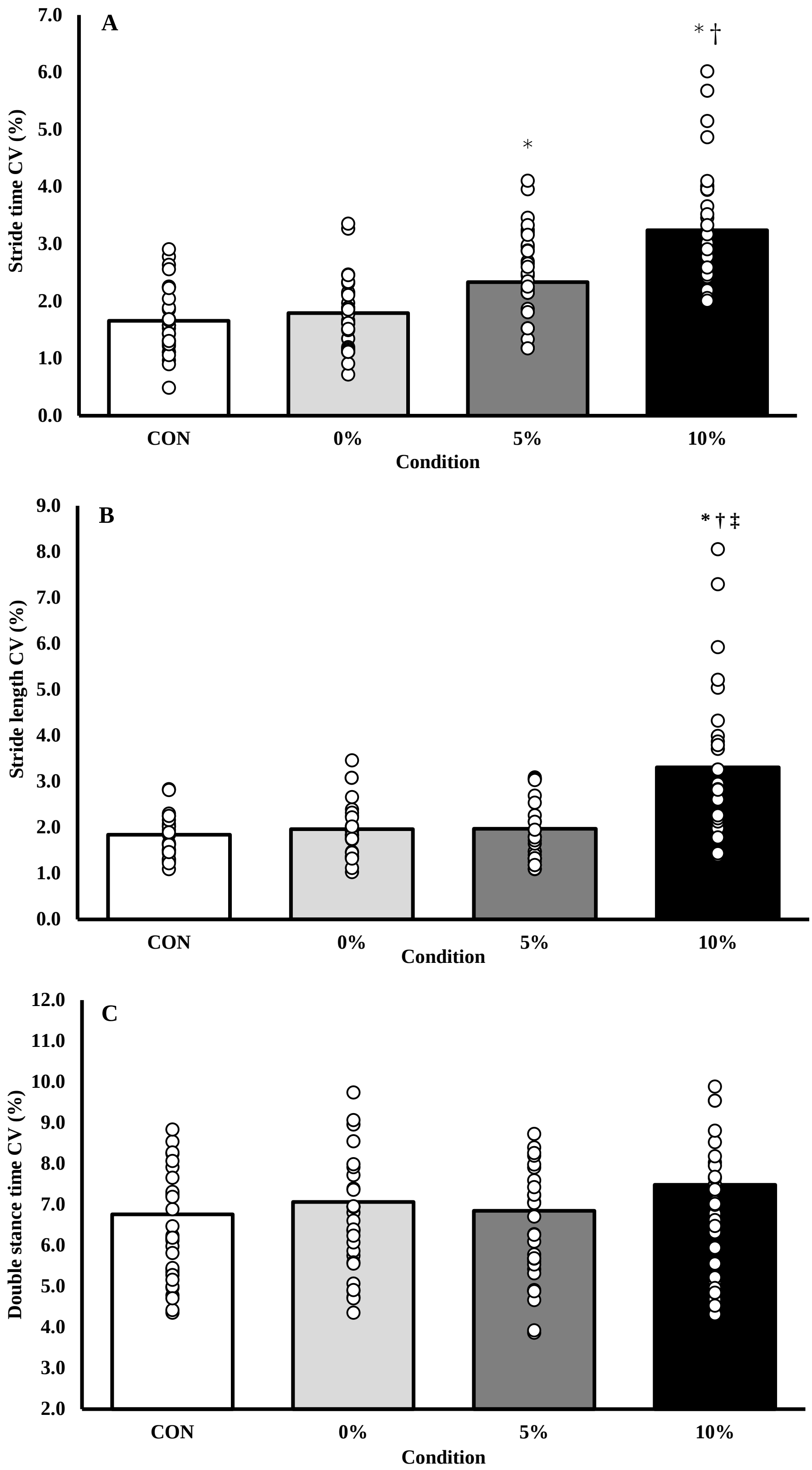

Fig. 3 Mean and individual stride-to-stride variability in stride time (A), stride length (B) and double limb stance time (C) during comfortable walking. *Significantly different to CON. †Significantly different to 0%. ‡Significantly different to 5%. ($P < 0.05$). ○ Represent individual data.

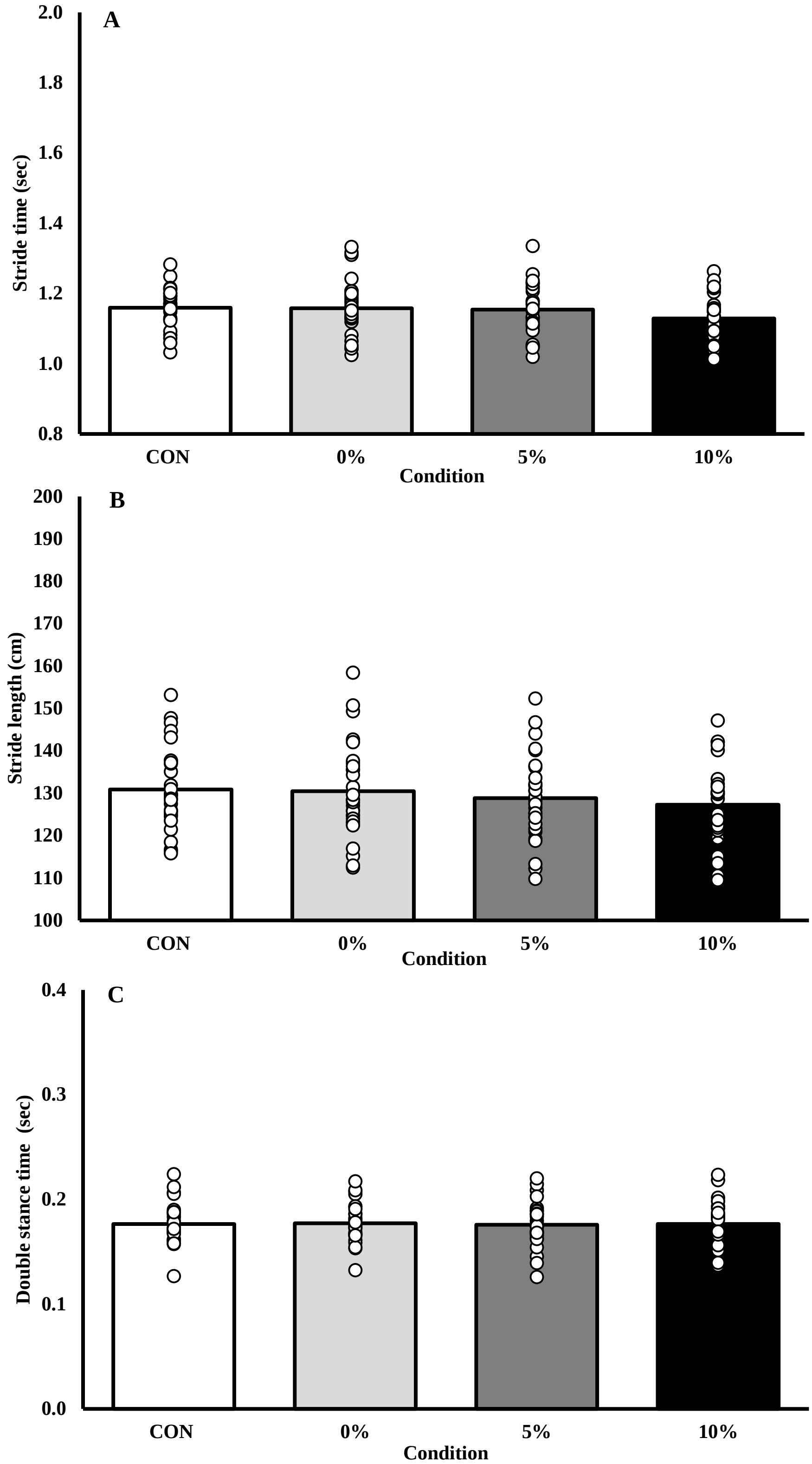

Fig. 4 Mean and individual stride time (A), stride length (B) and double limb stance time (C) during comfortable walking.

Table 1: Physiological and perceptual responses to walking during each load condition

	CON	0%	5%	10%
Heart rate (beats·min^{-1})	90 ± 15	93 ± 22	98 ± 15*	107 ± 17*†‡
% Age predicted HR_{MAX}	48.5 ± 8.0	50.1 ± 12.0	52.7 ± 8.5	57.5 ± 9.6
RPE_C	7 ± 2	8 ± 2*	8 ± 2*	10 ± 2*†‡
RPE_L	7 ± 1	7 ± 1	8 ± 2*	11 ± 2*†‡

*Significantly different to CON. †Significantly different to 0%. ‡Significantly different to 5%. ($P < 0.05$).