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Asymmetrical load-carrying while stepping down a curb in young adults

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ABSTRACT

Background: Asymmetrical load-carrying while walking requires modifications in joint forces to compensate the extra mass and ensure body stability, particularly when the environment is uneven, such as with a curb. Carrying a bag with one hand (dominant or non-dominant) may constrain the movement of the arm, altering the interlimb coordination of the upper limbs. Prior studies did not show changes in interlimb coordination when a light load was attached to the wrist, but the use of a bag to carry the load can be potentially disturbing since exaggerated movements of the bags may compromise balance. In this case, changes in interlimb coordination would be expected to minimize bag movements. However, it is not clear if these changes in interlimb coordination would be sufficient to affect the curb negotiation task.

Research question: We investigated the effect of asymmetric load-carrying using different bag types with the dominant and non-dominant hands on upper limb coordination and walking adaptations in a curb negotiation task in young adults.

Methods: Seventeen young adults walked and stepped down a curb while carrying a bag with 7% of their body mass. The experimental conditions were to walk without the bag, carrying the bag (with and without strap) using the dominant and non-dominant hand.

Results: Carrying the bag reduced the anti-phase pattern and increased the right or left shoulder phases, depending on the side used to carry the bag. It means that the limb that carried the load almost did not move while stepping down the curb. Load transportation did not influence foot-curb negotiation variables.

Significance: Our study indicates that a mild load and the bag influenced the interlimb coordination of the upper limbs. Despite that, young adults compensated for the disturbance caused by the load carriage and did not compromise the curb negotiation task.

1. Introduction

Walking is a daily performed action requiring the rhythmic activation of many muscles to produce the coordinated movement of body segments. The maintenance of dynamic stability and the ability to adapt the locomotion pattern to meet task and environmental demands are other essential requirements for successful locomotion [1,2]. Interestingly, young adults fall more often than previously thought [3]. The fall incidence was 52%, and 58% of these falls occurred during walking. Moreover, 25% of the falls were due to stumbling, and 14% of them happened when stumbling on curbs/steps.

The dimensions of mobility consider several aspects, including the external physical load and the terrain's characteristics [4]. After shopping, it is common to carry a bag and step down a curb in the way of locomotion. Stepping down a curb is a challenging task because it

demands careful control of foot trajectory to avoid contacting the edge of the curb as well as foot placement control to help to manage the lowering and horizontal displacement of the center of mass (COM) [5,6]. Past research has shown that young adults reduce their walking speed when stepping down a stairway while holding a cup of water [7]; in addition, changes in the interlimb coordination were also observed due to the imposed restriction on the upper limb motion. The anti-phase coordination pattern of the upper arms during walking is important for stability maintenance because it balances the trunk angular momentum around the vertical axis [8,9]. Donker et al. [10] did not find a change in upper limb coordination when walking and carrying a fixed 1.8-kg load attached to one of the wrists. Similarly, Matsuo et al. [11] showed that carrying a fixed mass of 3- or 8-kg did not affect the lower limb coordination when walking on a flat surface, but they observed compensatory adjustments at proximal joints. Carrying the load increased

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the hip abduction torque of the contralateral limb and decreased the torque in the ipsilateral hip.

Although Donker et al. [10] showed that carrying a load did not affect the coordination between upper limbs, the magnitude of the load and the way it was transported may have facilitated the task and minimized possible changes in coordination. Besides the mass, the characteristics of the load may represent an additional disturbance to the human walking. When carrying a bag with a strap, the movement of the bag is less predictable, and then more difficult to control compared to a bag without a strap (using a handle to hold it) since the load is close to the hand. Then, comparing these two types of bags will contribute to our understanding of the effect of asymmetrical load-carrying on walking coordination and control. Another aspect that can affect load carriage is the hand used to hold the bag. There are differences in perception and control mechanisms when using the dominant or non-dominant arms, which results in less advanced movement organization [12,13]. The use of the non-dominant hand may increase the challenge of the asymmetrical load-carrying.

We investigated the effect of an asymmetric load-carrying using different bag types with the dominant and non-dominant hands on upper limb coordination, global (COM) and local (foot-curb) movement adaptations in a curb negotiation task in young adults. Since we used a heavier load than Donker et al. [10] and a bag to carry the load, we expected to find a reduction in the anti-phase pattern for the right-left shoulder interlimb coordination. We examined only the shoulder coordination because this is a proximal joint that is mainly responsible for the anti-phase coordination mode observed during walking [14]. However, we did not expect changes in the foot-curb negotiation variables as Matsuo et al. [11] found compensatory adjustments at hip level when carrying a load. After all, carrying the load with the non-dominant limb has a potential to compromise the performance of the walking task because we are usually less skilled with this limb, which may increase the demands of the control system. Then, we hypothesized that body stability (assessed by the COM amplitude and velocity) would be compromised (i.e., increase in COM amplitude and velocity) during curb descent.

2. Methods

2.1. Participants

Seventeen healthy young adults (9 men, 8 women, 20.4 ± 2.2 years, 1.71 ± 0.08 m, 68.8 ± 10.6 kg) participated in this study. All participants exhibited right dominance for both upper and lower limbs [15,16]. The local ethics committee approved all procedures.

2.2. Procedures

We measured the grip strength for both dominant and non-dominant limbs [17]. Thirty-nine markers were placed on specific anatomical landmarks (*Plug-in-Gait Full Body* model) and were tracked by an 8-camera motion capture system (MX-T40S, Vicon, Oxford, UK), sampled at 100 Hz. Additionally, a marker placed on the upper right border of the curb identified its spatial location for further analyses.

For each participant, we adjusted the starting position located six steps before the curb. Participants walked at their preferred pace, stepped down the curb (0.16-m high), and continued walking to the ending position of the pathway (Fig. 1A). The leading limb for the curb descent was always the dominant one (right leg). The experimental conditions were to walk without the bag (no-load condition), carrying the bag (with and without strap) using the dominant and non-dominant hand. In both bag conditions, the participants took in one hand a bag filled with weights corresponding to 7% of their body mass. For the bag with a strap, participants held on the middle portion of the strap, whereas for the bag without the strap, they held on a piece of wood fixed to the border of the bag (Fig. 1B). In both bag conditions,

participants used the dominant and non-dominant hands. Participants performed three trials for each condition, totaling 15 randomized trials.

2.3. Data analysis

We digitally filtered the three-dimensional coordinates of each marker using a low-pass 4th-order Butterworth filter with a 6-Hz cut-off frequency. We computed the following foot-curb negotiation variables for both leading and trailing limbs as illustrated in Fig. 1C–D: stride length, duration and speed, foot-curb horizontal distance, curb-foot horizontal distance, vertical toe clearance, and horizontal heel clearance. The thirty-nine body markers defined a 15-body segment model used to compute the COM position based on Winter's [18] anthropometric parameters. COM velocity corresponded to the first derivative of the COM position (central difference procedure). The following COM variables were calculated only for the leading limb since some markers were missing when the trailing limb was stepping down the curb: medial-lateral (ML) amplitude (difference between the COM position at foot-contact of the trailing limb before the curb and the COM position at foot-contact of the leading limb after the curb), peak velocity in all three directions (maximum velocity observed in the interval between foot-contact of the trailing limb before the curb and foot-contact of the leading leg after the curb), COM velocity at foot contact after the curb, and the COM height at foot contact after the curb.

The Nexus software (Vicon) computed the right and left shoulder flexion-extension angles (Fig. 2A). We applied the vector coding technique [19–21] to identify the coordination patterns between left and right shoulder movements during the leading limb stride. Fig. 2B–C illustrates graphically the step-by-step to determine the coordination patterns. Based on the angle-angle diagrams for each consecutive pair of shoulder angles, the coupling angle (γ) was computed as the angle subtended from a vector adjoining two successive time points relative to right horizontal (see insert on Fig. 2B) [21]:

$$\gamma_{j,i} = \tan^{-1} \left(\frac{y_{j,i+1} - y_{j,i}}{x_{j,i+1} - x_{j,i}} \right)$$

where y and x represent the coordinates of the ordinate and abscissa of the angle-angle diagram (Fig. 2B).

The coupling angle quantifies the relative motion between two joints and allows the identification of four coordination patterns [14,19,21]: anti-phase, in-phase, right shoulder phase, and left shoulder phase (Fig. 2C). For the right shoulder phase, while the right shoulder moves, the left shoulder does not (or minimally) move and vice-versa for the left shoulder phase. The anti- and in-phase coordination patterns represent movements in both joints performed in the opposite and same direction, respectively. We classified the coupling angles into specific bins of 45° [19] (see Fig. 2 for details). When the coupling angle was 45° or 225° , the joints were in-phase; when the coupling angle was 135° or 315° , the joints were anti-phase; when the coupling angles was 0° or 180° , it characterized the right shoulder phase; and when the coupling angle was 90° or 270° , it defined the left shoulder phase. Based on these bins, we calculated the percentage of each coordination patterns in each stride.

2.4. Statistical analysis

For all variables, the mean of the three trials was used in the statistical analyses. For the foot-curb and gait variables, we used three MANOVA, combining the dependent variables of the leading and trailing limbs. The first MANOVA was one-way (load [no load, bag with strap - dominant and non-dominant hands]), with repeated measures. The second MANOVA was similar to the first one, but we used the bag without strap instead. The third MANOVA was two-way (side [dominant and non-dominant] x bag type [with and without the strap]), with repeated measures for both factors. For the COM variables, we used the

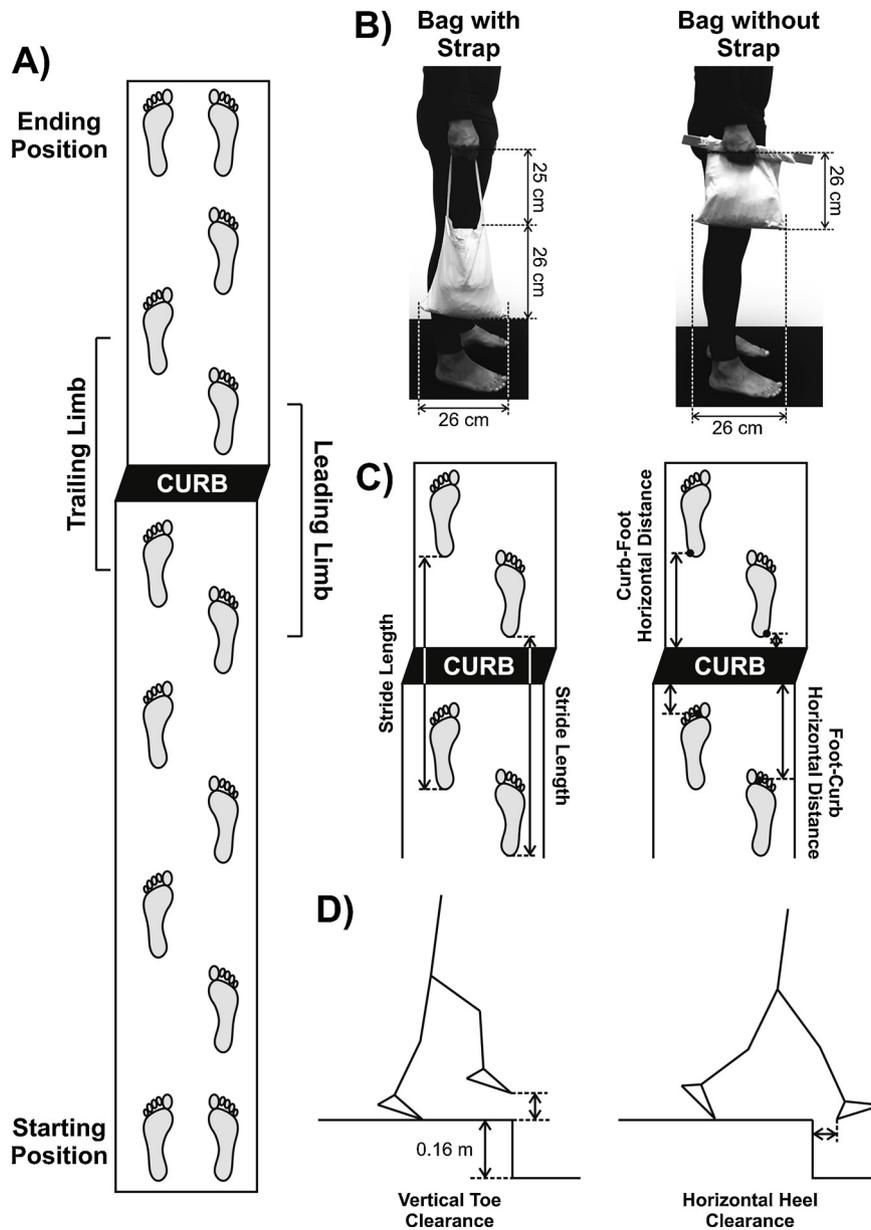


Fig. 1. A) Illustration of the whole pathway showing the starting position, curb, and ending position. It also identifies the leading and trailing limb in the curb descent phase. B) Pictures of a participant holding the two bags used in the present study and their respective dimensions. C) Illustration of the spatial-temporal and foot-curb variables. D) Diagram of the vertical and horizontal clearance measures.

same statistical models, but instead of running MANOVA, we carried out ANOVA since these variables were computed only considering the leading limb. For the coordination analysis, we used the same three MANOVA models explained above, but with the coordination modes (right/left shoulder, in-phase, and anti-phase) as the dependent variables. Univariate analyses followed each MANOVA. Post-hoc analyses with Bonferroni adjustments were carried out when main or interaction effects were identified. We also reported the partial eta-squared (η^2) for the MANOVA and univariate tests. The significance level was set at $p \leq 0.05$.

3. Results

The grip strength was higher for the dominant ($363.8 \pm 90.2\text{N}$) than for the non-dominant ($324.6 \pm 89.2\text{N}$) hand ($p \leq 0.0001$). The mean 7% load was equal to $4.8 \pm 0.7\text{kg}$.

3.1. Right-left shoulder coordination

The one-way MANOVA for both bag types exhibited a main effect of load (with strap: $p \leq 0.0001$, $\eta^2 = 0.934$ | without strap: $p \leq 0.0001$, $\eta^2 = 0.908$). The univariate tests showed that the effect was present in three coordination modes for both bag types (With Strap - right shoulder: $p \leq 0.0001$, $\eta^2 = 0.921$; left shoulder: $p \leq 0.0001$, $\eta^2 = 0.941$; anti-phase: $p \leq 0.0001$, $\eta^2 = 0.910$ | Without Strap - right shoulder: $p \leq 0.0001$, $\eta^2 = 0.922$; left shoulder: $p \leq 0.0001$, $\eta^2 = 0.914$; anti-phase: $p \leq 0.0001$, $\eta^2 = 0.857$). In both bag conditions, carrying the bag with the dominant hand increased the left shoulder phase and decreased the right shoulder phase compared to both the no-load and the non-dominant hand ($p \leq 0.001$, Fig. 3) and vice-versa for carrying the load with the non-dominant hand ($p \leq 0.019$). Moreover, carrying both bag types with either the dominant or the non-dominant hand reduced the anti-phase pattern compared to the no-load condition ($p \leq 0.0001$).

The two-way MANOVA revealed main effects of side ($p \leq 0.0001$,

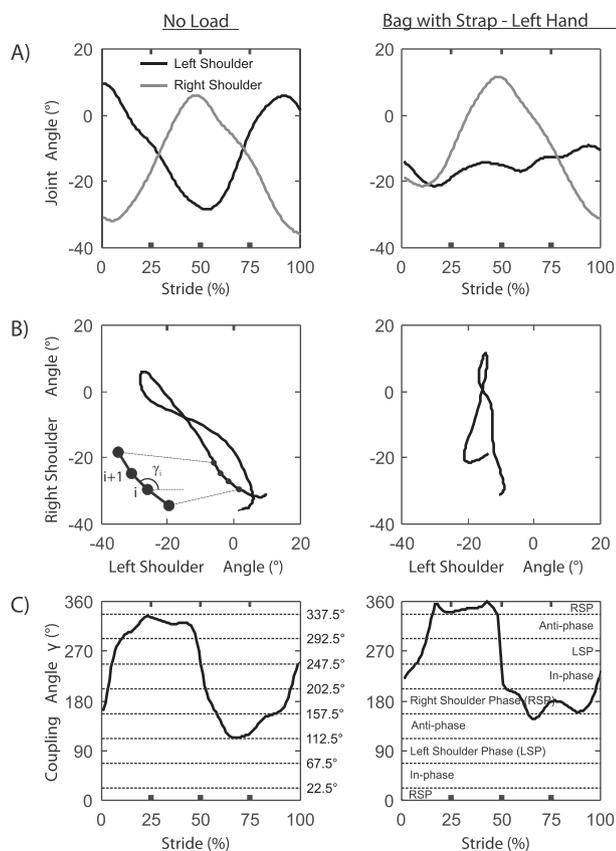


Fig. 2. A) Right and left joint angles during the leading limb stride (curb descent). The data is windowed from right heel-contact before the curb to right heel-contact after the curb. B) Angle-angle diagrams of the right and left shoulder joint angles shown in A. The insert illustrates four data points and the graphical example of only one coupling angle (γ_i), considering i and $i + 1$ pair of data points. The coupling angle was computed for each successive pair of data points throughout the whole gait cycle. C) Coupling angle for the data shown in B. The horizontal dashed lines indicates the intervals used to identify the four coordination patterns: right shoulder phase ($0^\circ \leq \gamma < 22.5^\circ$; $157.5^\circ \leq \gamma < 202.5^\circ$; $337.5^\circ \leq \gamma \leq 360^\circ$), in-phase ($22.5^\circ \leq \gamma < 67.5^\circ$; $202.5^\circ \leq \gamma < 247.5^\circ$), anti-phase ($112.5^\circ \leq \gamma < 157.5^\circ$; $292.5^\circ \leq \gamma < 337.5^\circ$), and left shoulder phase ($67.5^\circ \leq \gamma < 112.5^\circ$; $247.5^\circ \leq \gamma < 292.5^\circ$). The graphs in the left column are from one trial of the no-load condition and the graphs in the right column are from one trial of the bag with strap condition (left hand).

$\eta^2 = 0.985$) and bag type ($p = 0.049$, $\eta^2 = 0.496$), and an interaction between these two factors ($p = 0.005$, $\eta^2 = 0.658$). For the main effect of side, the univariate tests showed that the effect was present in both right and left shoulder phases ($p \leq 0.0001$, $\eta^2 = 0.975$ and $\eta^2 = 0.965$, respectively), and in the in-phase pattern ($p = 0.041$, $\eta^2 = 0.236$). The right shoulder phase was higher for the non-dominant ($71.6 \pm 2.4\%$) than for the dominant limb ($5.1 \pm 0.8\%$), whereas the left shoulder phase presented the opposite result (dominant: $73.3 \pm 2.8\%$ | non-dominant: $7.6 \pm 1.7\%$). The in-phase coordination pattern was slightly higher for the non-dominant ($13.7 \pm 1.1\%$) than for the dominant limb ($11.1 \pm 1.1\%$). The effect of bag type appeared only in the anti-phase coordination pattern ($p = 0.004$, $\eta^2 = 0.419$). When carrying the bag without strap, the anti-phase pattern was larger ($10.9 \pm 1.2\%$) than when carrying the bag with strap ($6.7 \pm 0.6\%$). The interaction was observed only for the left shoulder phase ($p = 0.003$, $\eta^2 = 0.442$). For the dominant limb, the left shoulder phase was slightly higher for the bag with strap than for the bag without strap, but there was no difference for the non-dominant limb (Fig. 3).

3.2. Foot-curb negotiation variables

Foot-curb variables were not affected by any of the load carrying conditions (Table 1).

3.3. Center of mass variables

For the COM, the observed differences were restricted to limb differences when carrying the bag, except for the ML amplitude (Table 2). For the ML amplitude, the one-way ANOVA for the bag without strap showed a main effect of load ($p = 0.004$, $\eta^2 = 0.287$). Carrying the bag in either dominant or non-dominant hand reduced the ML amplitude compared to the no-load condition ($p \leq 0.039$). For the remaining variables, there was no difference between no load and load conditions. For the peak AP velocity and AP velocity at foot contact, the two-way ANOVA exhibited a main effect of side ($p = 0.002$ and $p = 0.043$, $\eta^2 = 0.453$ and $\eta^2 = 0.233$, respectively). The AP velocity increased when carrying the bag with the non-dominant limb compared to the dominant limb ($p < 0.04$).

4. Discussion

We investigated the effect of carrying a load using different bag types with the dominant and non-dominant hands on upper limb coordination, global (COM) and local (foot-curb) adaptations in a curb negotiation task in young adults. We found that carrying a load restricted the upper limb movement as shown by the increase in right and left shoulder phases when using the non-dominant and dominant hands, respectively. There was also a reduction in the anti-phase coordination pattern when carrying the bag. Despite these changes in upper limb coordination, the foot-curb negotiation variables were not affected by the presence of the load.

For the COM, there was a slight increase in the AP velocity when carrying either bag with the non-dominant hand. Although statistically significant, this increase should be seen with caution because of the small difference. Asymmetrical load-carrying with the non-dominant limb seems to affect body stability because the rise in COM velocity contributes to reducing the margin of stability during walking [22]. We speculate that the use of the non-dominant hand results in poor control [12,13] that could compromise the manipulation of the bag, which in turn contributed to accelerating the body forward. It is possible that this poor control would demand more attention, which could impair walking control.

Interestingly, the COM ML amplitude reduced when carrying the bag without strap compared to the no-load condition. When carrying the bag, participants used a more conservative control of COM ML amplitude and deliberately constrained their movement in the frontal plane to prevent the bag from bouncing off the leg when changing the COM direction from one support limb to the other. Participants were better able to constrain the COM ML amplitude when using the bag without a strap.

4.1. Carrying a load modified the upper limb coordination

The use of the vector coding allowed us to identify and quantify the coordination modes present within the gait cycle [19,21]. Based on a prior study, we expected the anti-phase pattern to be predominant in the no-load condition (~60% of the gait cycle), followed by the right/shoulder phases (~15% each), and the in-phase pattern rarely present (< 10%) [14]. Our results for the no-load condition were close to these values, even though our participants stepped down a curb.

Carrying a mild load changed the upper limb coordination of the bag condition. The presence of the load increased the phase of the shoulder contralateral to that of the load, whereas the phase of the ipsilateral shoulder decreased. This finding means that the limb that carried the load practically did not move while stepping down the curb.

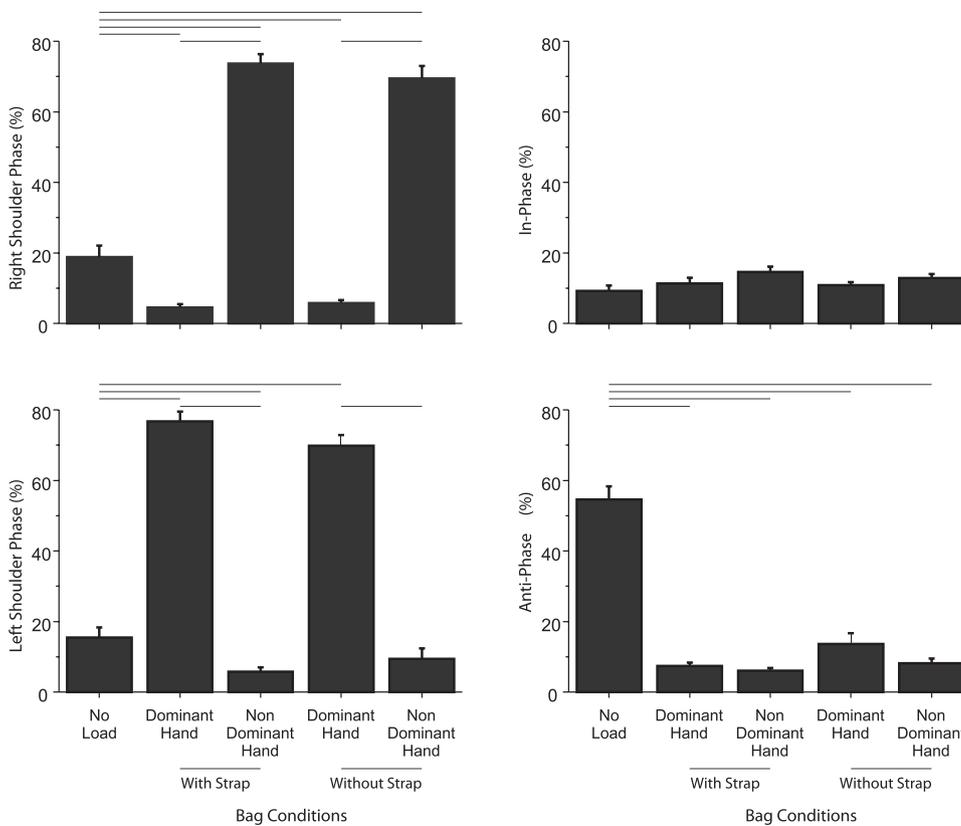


Fig. 3. Mean and standard error of the mean of the frequency of occurrence (%) of right shoulder (top left), in-phase (top right), left shoulder (bottom left), and anti-phase (bottom right) coordination patterns for flexion/extension movements for the coupling between right and left shoulder in all experimental conditions. Note that the sum up of all four phases for each condition equals approximately 100%. It is approximate because the values reported in the graphs represent the average value across participants. Horizontal lines indicate pairwise differences between experimental conditions ($p \leq 0.02$).

Table 1

Mean and standard error of the mean for the spatial-temporal gait parameters and the foot-curb variables for both leading and trailing limbs in all experimental conditions.

Variables	Conditions				
	No Load	Bag Without Strap		Bag With Strap	
		Non-dominant Limb	Dominant Limb	Non-dominant Limb	Dominant Limb
Stride Length (m)					
Leading	1.38 (0.03)	1.38 (0.05)	1.38 (0.04)	1.37 (0.05)	1.37 (0.03)
Trailing	1.46 (0.04)	1.42 (0.04)	1.42 (0.03)	1.41 (0.04)	1.43 (0.03)
Stride Duration (s)					
Leading	1.16 (0.03)	1.16 (0.03)	1.15 (0.03)	1.13 (0.04)	1.17 (0.02)
Trailing	1.08 (0.02)	1.07 (0.02)	1.07 (0.01)	1.07 (0.01)	1.07 (0.02)
Stride Speed (m/s)					
Leading	1.21 (0.03)	1.19 (0.04)	1.21 (0.03)	1.19 (0.04)	1.18 (0.03)
Trailing	1.36 (0.04)	1.32 (0.04)	1.33 (0.03)	1.32 (0.04)	1.34 (0.03)
Foot-Curb Horizontal Distance (m)					
Leading	0.88 (0.02)	0.89 (0.03)	0.90 (0.03)	0.92 (0.02)	0.90 (0.04)
Trailing	0.22 (0.02)	0.22 (0.01)	0.21 (0.01)	0.23 (0.01)	0.23 (0.02)
Vertical Toe Clearance (m)					
Leading	0.07 (0.003)	0.08 (0.003)	0.08 (0.003)	0.08 (0.003)	0.08 (0.003)
Trailing	0.08 (0.003)	0.09 (0.004)	0.08 (0.004)	0.08 (0.004)	0.08 (0.005)
Horizontal Heel Clearance (m)					
Leading	0.30 (0.02)	0.31 (0.02)	0.31 (0.02)	0.32 (0.01)	0.32 (0.02)
Trailing	0.50 (0.02)	0.48 (0.02)	0.49 (0.02)	0.49 (0.02)	0.48 (0.02)
Curb-Foot Horizontal Distance (m)					
Leading	0.36 (0.02)	0.33 (0.02)	0.34 (0.02)	0.34 (0.01)	0.34 (0.02)
Trailing	1.09 (0.04)	1.04 (0.04)	1.06 (0.02)	1.06 (0.02)	1.05 (0.03)

Simultaneously, as hypothesized, there was a reduction in the anti-phase coordination pattern with the load. This anti-phase pattern is known to help compensate for the trunk angular momentum [8,9]. It is possible that the load by itself could have been enough to balance the trunk angular momentum. The restriction of movement of the upper limb carrying the load could have been sufficient to dampen any perturbation resulting from the asymmetrical load carriage.

Donker et al. [10] found no change in upper limb coordination during walking with the addition of light load (1.8 kg) on the wrist. Two aspects can explain these inconsistent results. First, the load was heavier in the present study (~ 3 times) than in Donker et al.'s study leading to a physical constraint for moving the arm. To keep the typical arm movement with this load would potentially result in increasing energetic cost. Second, the use of a bag changes the dynamics of the

Table 2
Mean and standard error of the mean for the center of mass variables in all experimental conditions.

Variables	Conditions				Differences	
	No Load	Bag Without Strap		Bag With Strap		
		Non-Dominant Hand	Dominant Hand	Non-Dominant Hand		Dominant Hand
ML Amplitude (m)	0.21 (0.02)	0.15 (0.02)	0.16 (0.02)	0.17 (0.02)	1-way ANOVA (Bag Without Strap): No Load > Dominant and Non-Dominant (p = 0.039)	
Height at Foot Contact (m)	0.79 (0.01)	0.79 (0.01)	0.79 (0.01)	0.79 (0.01)	No Effect	
Peak ML Velocity (m/s)	0.111 (0.004)	0.114 (0.006)	0.106 (0.006)	0.113 (0.007)	No Effect	
Peak AP Velocity (m/s)	1.52 (0.044)	1.56 (0.042)	1.52 (0.042)	1.55 (0.040)	2-way ANOVA: Non-Dominant > Dominant (p = 0.002)	
Peak Vertical Velocity (m/s)	0.36 (0.032)	0.34 (0.027)	0.31 (0.028)	0.29 (0.037)	1-way ANOVA (Bag Without Strap): Non-Dominant > Dominant (p = 0.034)	
ML Velocity at Foot Contact (m/s)	0.044 (0.007)	0.053 (0.008)	0.038 (0.008)	0.049 (0.010)	No Effect	
AP Velocity at Foot Contact (m/s)	1.38 (0.042)	1.40 (0.040)	1.37 (0.040)	1.39 (0.035)	No Effect	
Vertical Velocity at Foot Contact (m/s)	-0.60 (0.024)	-0.61 (0.022)	-0.61 (0.024)	-0.57 (0.051)	2-way ANOVA: Non-Dominant > Dominant (p = 0.043)	

task compared to the mass attached to the wrist. The fact that the load was distal relative to the hand may generate random and disturbing movements if the hand is moving back-and-forth. Thus, the combination of load amount and bag characteristics may have been responsible for changing the upper limb coordination in the present study.

4.2. Carrying a load did not affect the foot-curb negotiation variables

Despite the changes in upper limb coordination, the walking adaptations to deal with the curb were unaffected. The appropriate control of foot placement and its trajectory are necessary conditions to avoid stumbling in the edge of the curb. Thus, young adults were able to compensate for the asymmetrical load-carrying, as shown by Matsuo et al. [11]. They may have prioritized the control of these parameters to ensure whole-body stability. On the other hand, our results contradict the findings of Madehkhaksar and Egges [7] who showed that carrying a cup of water reduced the walking speed when stepping down a staircase with four steps. Their upper task has a higher accuracy demand than ours, and this aspect could be sufficient to constrain the walking task. Additionally, the fact that their participants performed the task in a staircase may also influence since the demand is higher than walking down one step as in the present study.

5. Conclusion

Asymmetrical load-carrying changed the coordination pattern by reducing the movement of the limb with the load and by diminishing the anti-phase coordination pattern. Although these changes in upper limb coordination have the potential to compromise body stability, the foot-curb negotiation was unaffected by the load carrying task. Then, healthy young adults carrying a moderate load were able to compensate for the disturbance caused by the load carriage and did not compromise the curb negotiation task.

Declaration of Competing Interest

The authors declare that they have no conflict of interest.

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