The effects of (a)symmetric hand-held loads on gait performance in young males and females.

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Abstract

Hand-held loading is a common part of daily movement. However, its effects on gait control in young adults remain incompletely understood. Previous work suggests that arm swing, load placement, and sex may each influence gait performance, but how these factors interact during hand-held loading has not been systematically examined. Therefore, this study investigated how unilateral and bilateral hand-held loads affect spatiotemporal gait parameters, gait variability, and whole-body kinematics in healthy young adults, and whether males and females show different adjustment strategies under load. Thirty participants walked on a treadmill under three conditions: natural walking, unilateral loading at 10% body mass, and bilateral loading at 10% body mass. Whole-body 3D kinematic data were collected using a 10-camera Vicon motion-capture system, and spatiotemporal metrics and variability (CoV) were derived from heel-strike–based gait segmentation. Joint range of motion (ROM) and ROM variability were calculated for the shoulders, torso, and lower limbs.

Hand-held loading mainly influenced gait on the non-dominant side. Participants walked with shorter stride length, longer stride time, and slower stride speed when carrying a load. Shoulder swing amplitude and trunk rotation decreased across loading conditions. These changes suggest that participants relied on upper-body adjustments to maintain stability. Additionally, certain loading conditions increased gait variability, including larger left stride-time variability under bilateral loading and increased variability in shoulder and trunk rotation. Although males and females showed similar joint-angle adjustments under most conditions, some variability measures still showed sex-related differences.

Overall, this study demonstrates that hand-held loading modifies upper-body control strategies during walking and places additional demands on temporal and kinematic stability. These findings provide a foundation for future work examining how hand-held loading influences gait in older adults and clinical populations.

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Statement of Contribution of Collaborators

I, Jordan Yu, hereby declare that I am the sole author of this Master's thesis. The conception and design of this study were led by myself, with guidance and feedback from my supervisor, Dr. Julie Nantel. Dr. Nantel also provided valuable support during the early stages of planning the research objectives. I also received insightful feedback and support from my advisory committee, Dr. Cressman and Dr. Clouthier, particularly during the development of the research plan and the interpretation of results.

Participant recruitment was completed with the assistance of Dr. Nantel. Data collection was conducted by me in collaboration with members of our laboratory, including Elise, Ensieh, Ayham, and Shahab. I completed all data processing, analysis, and statistical procedures. I wrote this thesis independently, and Dr. Nantel provided academic guidance, feedback, and editorial corrections throughout the writing process.

I confirm that this thesis represents my original work and reflects my own contributions to the research design, data analysis, interpretation, and manuscript preparation.

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Definition of Terms

3D	Three-dimensional
ANOVA	Analysis of variance
BM	Body Mass
AP	Anterior–posterior
ML	Medio-lateral
VT	Vertical
CoM	Center of Mass
CoV	Coefficient of Variation
GG	Greenhouse–Geisser correction
ROM	Range of Motion
SD	Standard Deviation
SE	Standard Error
HS	Heel Strike

1. Introduction

Walking is one of the most common and repetitive activities in daily life. Functional gait is recognized as a walking pattern that requires stable and balanced forward progression with minimal energy expenditure to complete the gait tasks needed for daily living (Perry & Burnfield, 2010). In contrast, gait dysfunction may reduce the efficiency, stability, adaptability, and safety of gait (Maki, 1997; Hausdorff, 2007). Gait stability and gait variability play crucial roles in walking. Gait stability refers to maintaining a functional gait and preventing falling despite internal or environmental perturbations (Bruijn et al., 2013). These perturbations are constantly present while walking (Bruijn et al., 2010). Gait variability refers to the natural fluctuations in walking patterns (e.g., stride length, stride time, and stride speed), and it is a measurable characteristic that indicates the consistency or irregularity of motor control during walking (Hausdorff, 2005; Lord et al., 2011). It is considered a key factor when assessing gait performance, and it has been connected to increased risks of falls (Herman et al., 2010; König et al., 2014).. However, the relationship between gait stability and variability, as well as the influence of sex as a contributing factor to gait variability and stability, remains only partially understood.

A previous study by our group indicated that altered arm swing and asymmetric walking were independent factors contributing to the variability in gait spatiotemporal characteristics and joint trunk motion (Bailey et al., 2022). While active arm swings (e.g. increased swing amplitude) have been shown to enhance gait stability, they also led to increased variability in foot placement, trunk spatiotemporal characteristics, and lower limb joint angles while reducing coordination between lower limbs (Bailey et al., 2022; Hill & Nantel, 2019). Therefore, gait variability and lower limb kinematics reflect the necessary adjustments to stabilize gait in young adults.

Daily locomotor activities often involve carrying external loads such as briefcases, purses, or grocery bags. However, external loads are recognized as one of the factors influencing gait stability (Hoolihan et al., 2023; Matsuo et al., 2008; Wang et al., 2021). Previous research on healthy young adults has shown that using weighted vests and single-strap bags as asymmetric torso loads affects lower limb coordination, increases contralateral hip abduction torque, and decreases ipsilateral hip torque (Hoolihan et al., 2023; Wang et al., 2021). While these findings are vital for better understanding the impact of asymmetric loads on gait, most studies have focused on loads applied to the torso without considering the effect of arm motion on gait stability. Therefore, as hand-held loads are a common method of carrying loads, further research on how arm motion affects gait performance is needed.

Additionally, results by Rowe et al. (2021) suggests that the females greater pelvic width compared to males leads to smaller hip and ankle joint angles during walking, specifically in the frontal plane. They suggest that, due to the significant magnitude and pattern differences in lower limb kinematics and kinetics between males and females, sex-specific analyses are crucial in gait research. Furthermore, Cho et al. (2021) conducted a long-term quantitative study to track the impact of sex differences on fall frequency and circumstances. The results indicated that the occurrence and severity of falls were greater in young females than in young males, and they mainly occurred during everyday walking. While previous research has demonstrated sex differences in gait performance, the interaction between sex and external loading on gait adaptations remains to be determined.

This project aims to understand the impact of holding symmetrical (bilateral) and asymmetrical (unilateral) loading on gait performance, with a focus on identifying potential variations between sex. This knowledge would provide better weight-loading strategies and be critical in developing tailored gait (re)training programs tailored to men and women.

2. Research objective

This research aims to investigate whether holding external loads symmetrically and asymmetrically affects gait performance and examine the impact of sex. This research will use dumbbells to simulate daily loading and ask (a) how hand-holding external loads could interfere with gait patterns, and (b) what role, if any, sex-specific variants play in this interference between external loads and gait performance. To do so, participants will walk under three different load conditions:

1. Natural walking, no load (baseline)
2. One-hand (unilateral) loading at 10% body mass (BM)
3. Both hands loading (bilateral) at 10% BM

These investigations will enable us to: 1.1: Evaluate the impact of hand-held external loads on gait performance. 1.2: Examine the influence of sex on these variables.

Dependent variables:

1) Spatial-temporal gait parameters
   i. Stride time
   ii. Stride width
   iii. Stride length
   iv. Stride speed
2) Spatial-temporal variability (Coefficient of Variation (CoV), (standard deviation/mean) * 100)
   i. Stride time CoV
   ii. Stride width CoV
   iii. Stride length CoV
   iv. Stride speed CoV
3) Kinematics (range of motion (ROM) in three planes of motion: mediolateral (ML), anteroposterior (AP), and vertical (VT) planes)
   i. Shoulder ROM
   ii. Torso ROM

iii. Lower limbs ROM

4) Kinematic variability (ROM CoV in three planes of motion: ML, AP, VT)

i. Shoulder ROM CoV

ii. Torso ROM CoV

iii. Lower limbs ROM CoV

3. Hypotheses

It is hypothesized that 1.1: Hand-held load increases gait variability, especially with a unilateral load. The upper limbs will perform compensatory movements to enhance gait stability. 1.2: Females will exhibit greater gait variability when walking with a(symmetric) load(s) compared to males.

4. Literature review

4.1. Gait in young adults

As walking is an essential activity of daily life, an efficient gait, characterized by a regular walking pattern and appropriate arm swing (such as amplitude and frequency), can enhance the overall quality of life (Brandes et al., 2008; Pontzer et al., 2009). A gait cycle, defined as the movement of a single limb from heel strike to heel strike, is typically used to study the walking pattern (Gage et al., 1995). Gait stability and variability are commonly used to describe gait performance. Gait stability is defined as the ability to maintain functional gait and prevent falls in the presence of perturbations (Bruijn et al., 2013). Gait variability refers to the natural fluctuations in walking patterns, including stride length, width, time, and speed (Hausdorff, 2005). Increased gait variability often suggests a decline in gait stability, indicating that an individual may exhibit more irregular or unpredictable walking patterns (Hausdorff, 2005; Lord et al., 2011); therefore, it has been associated with an increased risk of falls (Callisaya et al., 2011; Hausdorff, 2005). However, a certain amount of gait variability has also been associated with healthy motor flexibility, which enables individuals to adjust and adapt gait patterns according to environmental demands (Bartlett et al., 2007). In addition, various factors influence gait variability, including age, sex, external factors (e.g., carrying loads, perturbations), and arm swing motion (e.g., symmetry, amplitude). Therefore, further research is needed to better understand how these factors affect gait variability, which may reduce gait variability and increase gait stability, ultimately improving quality of life.

4.2. Sex difference in gait performance

Significant differences in gait patterns exist between males and females. As early as the 1960s, Murray et al. (1967) reported that males exhibit greater torso ROM in the frontal plane, while females show greater pelvic ROM in the sagittal plane. Bruening et al. (2015) found that females showed greater variability than males, as quantified by higher root mean square values in the pelvis and torso in sagittal plane motion. Additionally, females had significantly greater shoulder ROM while walking. Based on these joint-angle variations, the authors concluded that males and females use different control strategies during walking (Bruening et al., 2015).

Moreover, Rowe et al. (2021) found that in the frontal plane, females exhibit significantly smaller ranges of knee flexion-extension and internal-external rotation angles when walking compared to males. The authors suggested that the age of the participants and the choice of gait speed were most likely the main reasons for the differences. However, this differs from other studies, which have shown sex-related differences only in the transverse plane for the pelvis and hip ROM, and in the sagittal plane for ankle ROM (Bruening et al., 2015, 2020; Røislien et al., 2009). However, considering the higher incidence of falls and fall-related injuries among females, especially in older adults (Cho et al., 2021), it is necessary to determine the sex-specific gait strategies used to regulate gait performance.

4.3. Gait performance in different arm swing strategies

Arm swing is a natural movement in human gait, which can reduce net metabolic energy expenditure by 5-7.7% (Ortega et al., 2009; Umberger, 2008; Collins et al., 2009). Early gait studies reported that arm swing resulted from the movement at the torso and inertia rather than active motion (Weber & Weber, 1836). Some studies suggested that the upper body functions like a mass-damped system in a passive arm wing model (Soong and Dargush, 1997; Symans and Constantinou, 1999). However, other studies have confirmed that arm swing plays a crucial role in gait, involving both active and passive movement (Elftman, 1939; Collins et al., 2009; Ortega et al., 2009; Umberger, 2009; Kuhtz-Bushbeck & Jing, 2012; Goudriaan et al., 2014). Electromyography (EMG) data confirmed this, as it showed that arm swing is a complex action involving both passive body motion and active contraction of the upper limb muscles, which could be used to adjust gait stability (Kuhtz-Buschbeck & Jing, 2012).

Collins et al. (2009) found that arm swing during walking reduces angular momentum around the vertical axis and vertical ground reaction torque, thereby decreasing lateral swaying of the body and improving gait stability. Yang et al. (2015) also examined the relationship between arm swing and gait stability through center of mass (CoM) displacement. They found that restricting arm swing significantly increased vertical CoM displacement, which was interpreted as a reduction in gait stability, as participants compensated by shortening their foot contact time and stride length. Conversely, it has been demonstrated that effective arm swing can improve gait stability (Siragy et al., 2020; Bailey et al., 2022; MacDonald et al., 2022).

Many recent studies have explored the effects of different arm swing strategies during walking (Bailey et al., 2022; Hu et al., 2012; MacDonald et al., 2022; Nakakubo et al., 2014;

Punt et al., 2015; Wu et al., 2016; Siragy et al., 2020). Punt et al. (2015) investigated the relationship between arm swing and gait using four different arm swing strategies: (1) without instruction, (2) swinging arms in phase with each other, (3) swinging arms in phase with ipsilateral legs, and (4) performing arm swing with large amplitude. They found that gait was most stable with increased arm swing amplitude. However, studies by our group showed that while active arm swing benefits local trunk stability (measured by the Lyapunov exponent, which assesses sensitivity to small disturbances), it also increases variability in foot placement, spatiotemporal characteristics of the trunk, and stride variability in lower limb joint angles during stable conditions (Bailey et al., 2022; Hill & Nantel, 2019; Siragy et al., 2020), which suggests that young adults actively exploit lower limb redundancy to regulate trunk stability and gait performance (Hill & Nantel, 2019). Considering that arm swing contributes to the regulation of whole-body movement during walking, any deviation from natural arm motion may require compensatory adjustments in trunk and lower-limb kinematics. The adaptations could reflect increased demands on gait regulation and may influence overall gait performance. This remains to be investigated.

4.4. Gait performance while carrying external loads

When walking, people often carry external loads, such as a briefcase or grocery bags. In the literature, the methodology for assessing the effect of external loads on gait is generally divided into two types of loads: 1. weighted vests to distribute the weight evenly across the torso and 2. backpacks to simulate daily load-carrying situations. Weight placement simulates symmetric and asymmetric conditions to determine their effect on gait performance. Some research found that when carrying asymmetric external loads, the contralateral trunk flexes to compensate for the load imbalance, which alters gait performance (Fowler et al., 2006). Yogev-Seligmann et al. (2008) simulated asymmetric external loads with a single-strap messenger bag. They found that this condition significantly impacted lower limb coordination, which resulted in gait asymmetry.

A previous study examining the effects of hand-held and back loads on the gait cycle showed different gait strategies as a function of the load types. When hand-held loads (in the ipsi or contralateral hand) reached 20% of body weight, the duration of the stance phase decreased, and the duration of muscle activity for the contralateral gluteus medius and ipsilateral gastrocnemius, vastus lateralis, and semimembranosus was significantly prolonged. However, when walking with a back load between 20-50% of the body weight, the average duration of the swing phase significantly decreased, and the vastus lateralis' electromyographic activity increased to maintain gait stability in young adults (Ghori & Luckwill, 1985). Another study indicated that older adults, when descending stairs with asymmetric hand loads, adjusted the position of the CoM in the ML direction, with the trunk leaning toward the side without the load, and exhibited minimal shoulder movement of the load-carrying side to increase gait stability. The movement of both shoulders became more synchronized, contrary to unloaded or symmetrically loaded arm motion. The authors suggested that this was done to reduce the impact of the load on the movement at the torso and improve gait stability (Silva et al., 2020). Additionally, Matsuo et al. (2008) conducted a study in women carrying a hand-held bag. The authors reported increased contralateral hip abduction and decreased ipsilateral hip abduction when carrying asymmetric loads. As the load increased, participants further modified their trunk kinematics, as well as the contralateral arm, to maintain balance. However, a systematic investigation of whether (a)symmetrical hand-held weights affect arm swing strategies and gait stability has yet to be carried out.

Lastly, previous studies have shown significant sex differences in thigh-shank coupling while carrying external loads using a weighted vest in young adults. Females had larger coupling variability in their lower limbs than males (Hoolihan et al., 2023). However, this difference was observed at a relative load of 40% BM, and this study included only young adults. Therefore, further comprehensive research is required to understand the impact of sex on lower limb joint angles and gait performance when carrying hand-held loads.

5. Method

5.1. Participants

Thirty healthy young adults (15 males, 15 females) were recruited from the Ottawa area. Young adults were defined as participants aged 18–35 years, consistent with previous studies on gait and motor control. This study included two groups of participants (young males, young females) and three loading conditions (natural walking, single-hand load, and double-hand load). According to G*Power (Faul et al., 2007), the statistical analysis was a two-way repeated measures analysis of variance (ANOVA). The effect size was 0.25, indicating a medium effect (Cohen, 1988). The alpha error probability (α) was set at 0.05, while the power (1 - β error probability) was set at 0.80. The resulting sample size should be at least 28 participants. Therefore, this study recruited 15 participants in each group. Participants were excluded if they had had a musculoskeletal injury in the past six months or any chronic neurological or orthopaedic disorders. All participants provided written informed consent, and the study was approved by the local ethics committees.

5.2. Experimental procedure

The asymmetrical load was placed in the non-dominant hand in this study. People often use their non-dominant hand to lift or carry heavy loads, leaving their dominant hand free for more fine-motor tasks (e.g. unlocking a door). In this study, all participants were right-hand dominant; therefore, the load was always carried in the left hand. Accordingly, the results were presented as left and right rather than dominant and non-dominant sides. Each participant walked on a treadmill for five minutes in three loading conditions: 1.

natural walking without loading (baseline), 2. non-dominant hand loaded with 10% BM, and 3. both hands loaded with 10% BM. According to the American Physical Therapy Association's guidelines, appropriate daily hand loading should be controlled between 10-15% BM. Previous studies also used 10-20% BM loading to observe changes in locomotor behavior. Participants walked at their preferred speed and were allowed a 5-10-minute rest between trials. Each participant wore a full-body motion capture suit and was instrumented with a 57-marker set (Wilken et al., 2012; Sinitski et al., 2015). Whole-body three-dimensional (3D) kinematic data were collected using a 10-camera Vicon motion capture system (Vicon 2.10.2, Oxford, UK). Kinematic data were captured at 100Hz. Data were processed and labelled using Vicon Nexus (version 2.10.2). Participants wore their athletic shoes during the experiment.

5.3. Data analysis

Marker trajectories were used in Visual3D (C-Motion, Germantown, MD, USA) to create 3D model of joint movement (Collins et al., 2009; Wilken et al., 2012). Marker data were processed by Vicon Nexus (Nexus 2.10.2, Oxford, UK) and then exported to Visual3D v6 (C-Motion, Germantown, MD) for 3D kinematic analysis, including the calculation of ROM in the ML, AP, and VT planes of motion for the shoulders, torso, and lower limbs. Raw data were filtered at a 10 Hz cut-off frequency using a 2nd-order low-pass Butterworth filter (Winter, 2009).

The collected data were processed and analyzed using Matlab (Mathworks, Natick, MA) to produce spatiotemporal parameters, including step time, width, length, and walking speed. Spatiotemporal variability (step length, width, time) was measured by calculating the CoV ((standard deviation/mean) * 100), which showed the degree of variation relative to the mean value of each parameter.

The laboratory axes were aligned with X representing the anteroposterior (AP) direction, Y representing the mediolateral (ML) direction, and Z representing the vertical (VT) direction. Heel-strike (HS) events were identified from local minima in the heel-marker's vertical trajectory (LHEE, RHEE), following kinematics-only event detection approaches validated for treadmill and overground walking (O'Connor et al., 2007; Zeni, Richards, & Higginson, 2008).

Stride time was measured as the interval between consecutive ipsilateral HSs. Since the net AP displacement of the foot in the global frame is minimal on a motorized treadmill, we reported AP excursion per stride (a proxy for stride length) as the difference in the ipsilateral heel's X-coordinate between consecutive HSs; the corresponding excursion rate (proxy for speed) was calculated as excursion divided by stride time (Perry & Burnfield, 2010). Stride width was the ML distance between heel markers at the left HS (a toe-defined variant was also calculated), consistent with clinical gait analysis conventions (Perry & Burnfield, 2010). Metrics were computed separately for the left and right sides and summarized as mean ± standard error (SE).

Each condition consisted of a continuous 5-minute trial at the participant's self-selected speed. To minimize non-steady transients at the beginning and end, we analyzed only the central 3 minutes. The initial phase typically involves familiarization with belt speed, adjustments to the load configuration, and re-tuning of arm swing and trunk motion (Meyer et al., 2019); the terminal phase can include anticipatory behaviors related to trial termination, attentional drift, or early fatigue. Restricting analyses to the middle segment improves time-series stationarity, stride-time, and variability estimates that are more representative of steady-state gait, consistent with recommendations to assess gait stability and variability under steady-state conditions (Bruijn et al., 2013).

Previous research shows that spatiotemporal gait parameters vary systematically with body size. Therefore, normalization helps reduce between-subject variability and prevents confounding in group comparisons (Hof, 1996; Pinzone, Schwartz, & Baker, 2016). In this study, height-based normalization was applied to the stride length, stride speed, and step width by dividing each variable by the participant's standing height (cm). Normalized values were therefore expressed as mm/cm or (mm/s)/cm. This approach is commonly used to help separate body-size effects from control strategies (Donelan, Kram, & Kuo, 2001).

A two-way mixed-design repeated measures ANOVA was conducted using SPSS (IBM Corp. Released 2020. IBM SPSS Statistics for MacOS, Version 30.0. Armonk, NY: IBM Corp) to determine the effects of external loading and sex on the dependent variables. External loading was set as a within-subject factor with three conditions (Natural walking, one-hand loading at 10% BM, and both-hand loading at 10% BM), while sex (male and female) was a between-subject factor. The interaction effects were also examined to determine whether the external loading influenced the differences between sex groups. The significance level was set at $p < .05$ for all analyses, and a Bonferroni correction was applied for post-hoc pairwise comparisons, ensuring a more accurate interpretation of p-values. Because the within-subject factor "Load" has three levels, the repeated-measures ANOVA required the sphericity assumption (equal variances of all pairwise level differences). When Mauchly's test indicated a violation of sphericity, we reported Greenhouse–Geisser (GG) corrected p-values (with adjusted degrees of freedom) to control for Type I error inflation.

6. Results

6.1 Participant demographics

Thirty healthy young adults (15 males, 15 females; 19–33 years) participated in the study. Age did not differ by sex ($p = .335$), and males were taller and heavier than females (both $p < .001$; Table 1).

Table 1. Participant Characteristics

Variable	Male (n=15)	Female (n=15)	Total (N=30)	P values
Age (years)	25.5 ± 4.1 years (19–32)	26.9 ± 3.7 years (20–33)	26.2 ± 3.9 years (19–33)	0.335
Height (cm)	179.4 ± 7.7 cm (165–191)	162.7 ± 9.2 cm (148–178)	171.0 ± 11.9 cm (148–191)	<0.001
Weight (kg)	81.1 ± 10.5 kg (66.8–101.5)	57.1 ± 6.5 kg (45.9–70.0)	69.1 ± 15.0 kg (45.9–101.5)	<0.001

Note. Values are mean ± SD; parentheses indicate range (min–max). p values from Welch's t-test comparing males vs females

6.2 Spatial-temporal gait parameters

Table 2a shows the main effects of Load and the Load × Sex interaction on spatiotemporal gait parameters. Relative to baseline, both one-hand and two-hand loading led to shorter left stride length ($p = .002$), longer left stride time (GG $p = .008$), and slower left stride speed ($p < .001$). Right stride time showed a marginal effect (GG $p = .060$). Stride width (heel and toe) did not show a significant difference with load ($p \geq .274$) (Table 2a). A single interaction emerged for left stride speed ($p = .030$), indicating different speed adjustments across sexes under load (Table 2a).

Table 2a. Main effect of Load and Load × Sex interaction on spatiotemporal gait parameters

Variable	Baseline	One-hand loading	Two-hand loading	Main effect of Load (p)	Sex × Load (p)
Stride length L (mm)	345.7 ± 13.3	334.3 ± 13.2	331.2 ± 13.1	**.002 ****	.066 †
Stride length R (mm)	335.8 ± 13.8	350.1 ± 11.4	341.4 ± 12.3	0.249	0.775
Stride time L (s)	0.728 ± 0.008	0.733 ± 0.009	0.750 ± 0.011	**.008 ** (GG)**	0.563
Stride time R (s)	0.735 ± 0.011	0.713 ± 0.007	0.724 ± 0.009	.060 † (GG)	0.119
Stride speed L (mm/s)	482.6 ± 17.9	463.7 ± 18.5	452.0 ± 17.8	**<.001 *****	**.030 ***
Stride speed R (mm/s)	471.0 ± 22.2	500.7 ± 18.1	484.6 ± 20.8	0.153	0.942
Stride width Heel (mm)	33.0 ± 2.4	32.4 ± 2.0	32.3 ± 2.1	0.875	0.56
Stride width Toe (mm)	25.5 ± 1.3	26.4 ± 1.4	26.6 ± 1.2	0.274	0.746

Note. Values are estimated marginal means ± standard error (SE). †$p < .10$, *$p < .05$, **$p < .01$, ***$p < .001$. For measures with sphericity violations, Greenhouse–Geisser corrected p-values are reported.

Post-hoc tests following the main effects of Load are presented in Table 2b. For left stride length, both one-hand and two-hand loading showed smaller stride length compared to baseline (p = .037 and p = .006), with no difference between the two loading conditions (p = 1.000). For left stride time, two-hand loading was greater than baseline (p = .018). Differences between one-hand and two-hand loading were not significant after Bonferroni adjustment. For left stride speed, one-hand and two-hand were slower than baseline (p = .017 and p = .002), with no difference between the two loading conditions (p = .320)

Table 2b. Pairwise load comparisons for spatiotemporal gait parameters (Bonferroni-adjusted)

Variable	Baseline vs One-hand Δ (95% CI)	p-value	Baseline vs Two-hand Δ (95% CI)	p-value	One-hand vs Two-hand Δ (95% CI)	p-value
Stride length L (mm)	+11.34 [0.56, 22.12]	**0.037 ***	+14.52 [3.57, 25.46]	**0.006 ***	+3.18 [−6.39, 12.74]	1
Stride time L (s)	−0.006 [−0.018, 0.006]	0.638	−0.022 [−0.041, −0.003]	**0.018 ***	−0.016 [−0.035, 0.002]	.099 †
Stride speed L (mm/s)	+18.87 [2.80, 34.94]	**0.017 ***	+30.54 [10.07, 51.01]	**0.002 ***	+11.67 [−6.16, 29.49]	0.32

Note. Δ = mean difference (first − second condition) based on estimated marginal means. Positive values mean the first condition is larger. †p < .10, *p < .05.

Table 2c shows that Sex had significant effects on left stride length (p = .039), left stride speed (p = .036), and toe-based step width (p = .014). No sex effects were observed for right stride length, left/right stride time, right stride speed, or heel-based step width (all p ≥ .062). After height-based normalization of left stride length, left speed, and toe-based step width (mm or mm/s per cm), the sex effects were no longer significant (all p ≥ .116; see Appendix C, Table C1).

Table 2c. Main effect of Sex on spatiotemporal gait parameters

Variable	F(df=1,28)	p-value	Partial η²	Power
Stride length L (mm)	4.67	**0.039***	0.143	0.55
Stride length R (mm)	0.322	0.575	0.011	0.085
Stride time L (s)	0.484	0.492	0.017	0.103
Stride time R (s)	1.672	0.207	0.056	0.239
Stride speed L (mm/s)	4.839	**0.036***	0.147	0.565
Stride speed R (mm/s)	0.595	0.447	0.021	0.116
Stride width Heel (mm)	3.785	0.062	0.119	0.468
Stride width Toe (mm)	6.843	**0.014***	0.196	0.714

Note. Asterisks (*) show statistical significance at p < .05.

6.3 Spatial-temporal gait variability

Table 3a shows the main effects of Load and the Load × Sex interaction on spatiotemporal gait variability (CoV). Left stride time CoV increased with loading (GG p = .011). Toe width CoV was nominally lower with loading but did not reach statistical significance (p = .093).

Table 3a. Main effect of Load and Load × Sex interaction on spatiotemporal gait variability (CoV)

Variable	Baseline	One-hand loading	Two-hand loading	Main effect of Load (p)	Sex × Load (p)
Stride length L (CoV, %)	18.11 ± 1.18	19.04 ± 1.12	18.55 ± 0.88	0.427	0.104
Stride length R (CoV, %)	15.53 ± 1.21	15.45 ± 1.16	16.84 ± 1.33	0.283	0.962
Stride time L (CoV, %)	10.83 ± 1.00	11.73 ± 1.06	12.99 ± 1.04	**0.011* (GG)**	0.542
Stride time R (CoV, %)	9.53 ± 0.75	9.11 ± 0.67	9.79 ± 0.80	0.54	0.706
Stride speed L (CoV, %)	20.97 ± 1.48	20.95 ± 1.22	21.48 ± 1.05	0.819	0.608
Stride speed R (CoV, %)	20.55 ± 1.65	20.30 ± 1.51	21.79 ± 1.61	0.473	0.539
Stride width Heel (CoV, %)	41.20 ± 3.63	43.24 ± 3.64	44.95 ± 3.32	0.295	0.935
Stride width Toe (CoV, %)	38.66 ± 2.51	35.57 ± 2.54	34.76 ± 2.31	0.093 †	0.74

Note. Values are estimated marginal means ± standard error (SE). †p < .10, *p < .05, **p < .01, ***p < .001. For measures with sphericity violations, Greenhouse–Geisser corrected p-values are reported.

Table 3b shows the Bonferroni-adjusted post-hoc pairwise comparisons for load on spatiotemporal gait variability. For left stride time CoV, two-hand was higher than baseline (p = .012). The contrasts between baseline and one-hand (p = .127) and between one-hand and two-hand (p = .318) were not significant.

Table 3b. Pairwise load comparisons for spatiotemporal gait variability (CoV, Bonferroni-adjusted)

Variable	Baseline vs One-hand Δ (95% CI)	p-value	Baseline vs Two-hand Δ (95% CI)	p-value	One-hand vs Two-hand Δ (95% CI)	p-value
Stride time L (CoV, %)	−0.90 (−1.97, 0.18)	0.127	−2.16 (−3.91, −0.40)	**0.012 ***	−1.26 (−3.18, 0.66)	0.318

Note. Δ = mean difference (first − second condition) based on estimated marginal means. Positive values mean the first condition is larger. Adjusted p values use Bonferroni. *p < .05.

Sex showed a significant main effect on right stride length variability ($p = .032$). No sex differences were observed for left and right stride time, left stride length, left and right stride speed, or stride width at the heel or toe (all $p \geq .206$) (Table 3c).

Table 3c. Main effect of Sex on spatiotemporal gait variability (CoV)

Variable	F(df=1,28)	p-value	Partial η²	Power
Stride length L (CoV, %)	0.17	0.684	0.006	0.068
Stride length R (CoV, %)	5.08	**0.032***	0.154	0.586
Stride time L (CoV, %)	0.75	0.393	0.026	0.134
Stride time R (CoV, %)	0.42	0.52	0.015	0.096
Stride speed L (CoV, %)	0.14	0.711	0.005	0.065
Stride speed R (CoV, %)	1.68	0.206	0.057	0.24
Stride width Heel (CoV, %)	1.1	0.303	0.038	0.174
Stride width Toe (CoV)	0	0.969	0	0.05

Note. Asterisks (*) show statistical significance at $p < .05$.

6.4 Kinematics ROM

Load had significant main effects on left hip flexion, bilateral shoulder flexion, and torso kinematics (Table 4a). Specifically, left hip flexion differed across conditions ($p = .006$), shoulder flexion decreased with increasing load on both sides (right: $p < .001$; left: $p < .001$), and the torso showed reduced rotation ($p < .001$; GG-corrected), a slight decreased at one-hand loading and recovered at two-hand ($p = .001$), and slightly greater lateral bending during two-hand loading ($p = .006$). Trend-level effects were seen for right ankle angle ($p = .062$) and left hip adduction ($p = .056$). No load effects were found for right hip flexion, knee angles, left ankle angle, or right hip adduction (all $p \geq .129$). The only Load × Sex interaction was for left hip adduction ($p=.014$); all others were not significant (all $p \geq .129$; GG-corrected as needed).

Table 4a. Main effect of Load and Load × Sex interaction on kinematics ROM

Variable	Baseline	One-hand loading	Two-hand loading	Main effect of Load (p)	Sex × Load (p)
Hip flexion R (°)	40.72 ± 1.53	40.89 ± 1.48	41.01 ± 1.66	0.904	0.633
Hip flexion L (°)	39.90 ± 1.35	38.71 ± 1.25	40.62 ± 1.44	**0.006 ****	0.873
Knee angle R (°)	55.15 ± 1.87	55.16 ± 1.64	55.16 ± 1.89	1.000	0.175
Knee angle L (°)	54.67 ± 2.01	54.93 ± 1.96	54.68 ± 2.00	0.935	0.834
Ankle angle R (°)	28.55 ± 1.47	27.56 ± 1.46	28.09 ± 1.48	0.062 †	0.202
Ankle angle L (°)	27.50 ± 1.06	27.96 ± 1.14	28.27 ± 1.23	0.304	0.129
Hip adduction R (°)	16.36 ± 0.63	16.48 ± 0.65	16.20 ± 0.64	0.773	0.242
Hip adduction L (°)	15.19 ± 0.63	14.31 ± 0.54	15.17 ± 0.53	0.056 †	**0.014 ****
Shoulder flexion R (°)	21.26 ± 2.52	13.46 ± 1.46	5.62 ± 0.51	**<.001 *****	0.598 (GG=0.548)
Shoulder flexion L (°)	19.68 ± 1.88	5.51 ± 0.77	5.11 ± 0.40	**<.001 *** (GG<.001)**	0.797 (GG=0.697)
Torso extension (°)	5.83 ± 0.24	5.39 ± 0.23	5.92 ± 0.28	**0.001 ****	0.347
Torso bending (°)	9.87 ± 0.51	9.52 ± 0.45	10.42 ± 0.48	**0.006 ****	0.333
Torso rotation (°)	12.43 ± 0.68	8.62 ± 0.38	7.87 ± 0.44	**<.001 *** (GG<.001)**	0.926 (GG=0.843)

Note. Values are estimated marginal means ± standard error (SE). †p < .10, *p < .05, **p < .01, ***p < .001. For measures with sphericity violations, Greenhouse–Geisser corrected p-values are reported.

Table 4b shows the Bonferroni-adjusted post-hoc pairwise comparisons for load on joint ROM. For left hip flexion, one-hand loading was lower than two-hand loading (p = .005), and baseline did not differ from either loading condition (p ≥ .098). The shoulder flexion declined as the load increased on both sides: on the right, one-hand loading was lower than baseline, and two-hand loading was lower than one-hand loading (all p ≤ .015); on the left, one-hand loading and two-hand loading were both lower than baseline (both p < .001), with no difference between the two loading conditions (p = 1.000). For the torso, extension was smaller with one-hand loading than with two-hand loading (p = .001) and baseline did not differ from two-hand loading (p = 1.000); lateral bending was smaller with one-hand loading than with two-hand loading (p = .003); and rotation decreased from baseline to one-hand loading to two-hand loading (all p ≤ .002).

Table 4b. Pairwise load comparisons for kinematics ROM (Bonferroni-adjusted)

Variable	Baseline vs One-hand Δ (95% CI)	p-value	Baseline vs Two-hand Δ (95% CI)	p-value	One-hand vs Two-hand Δ (95% CI)	p-value
Hip flexion L (°)	1.19 (−0.16, 2.53)	0.098	−0.73 (−2.35, 0.89)	0.783	−1.91 (−3.33, −0.50)	0.005 **
Shoulder flexion R (°)	7.80 (1.30, 14.30)	0.015 *	15.64 (9.49, 21.78)	<.001 ***	7.84 (4.17, 11.51)	<.001 ***
Shoulder flexion L (°)	14.17 (9.50, 18.84)	<.001 ***	14.56 (9.62, 19.50)	<.001 ***	0.39 (−1.71, 2.49)	1
Torso extension (°)	0.44 (0.08, 0.80)	0.014 *	−0.10 (−0.53, 0.34)	1	−0.54 (−0.86, −0.21)	0.001 **
Torso bending (°)	0.35 (−0.30, 0.99)	0.559	−0.55 (−1.33, 0.23)	0.248	−0.90 (−1.51, −0.28)	0.003 **
Torso rotation (°)	3.81 (2.67, 4.94)	<.001 ***	4.56 (3.38, 5.73)	<.001 ***	0.75 (0.25, 1.25)	0.002 **

Note. Δ = mean difference (first − second condition) based on estimated marginal means. Positive values mean the first condition is larger. Adjusted p values use Bonferroni. *p < .05. , **p < .01, ***p < .001.

Sex did not show a main effect on joint ROM, with the left ankle angle being the smallest non-significant p-value ($p = .073$) (Table 4c).

Table 4c. Main effect of Sex on kinematics ROM

Variable	**F(df=1,28)**	**p-value**	**Partial η²**	**Power**
Hip flexion R (°)	0.631	0.434	0.022	0.12
Hip flexion L (°)	0.367	0.549	0.013	0.09
Knee angle R (°)	0.026	0.874	0.001	0.053
Knee angle L (°)	0.098	0.757	0.003	0.061
Ankle angle R (°)	0.349	0.559	0.012	0.088
Ankle angle L (°)	3.468	0.073	0.11	0.436
Hip adduction R (°)	0.305	0.585	0.011	0.083
Hip adduction L (°)	0.095	0.761	0.003	0.06
Shoulder flexion R (°)	1.179	0.287	0.04	0.182
Shoulder flexion L (°)	0.538	0.47	0.019	0.109
Torso extension (°)	2.615	0.117	0.085	0.345
Torso bending (°)	0.198	0.66	0.007	0.071
Torso rotation (°)	2.781	0.107	0.09	0.364

Note. Asterisks (*) show statistical significance at p < .05.

6.5 Kinematics ROM variability

Load had significant main effects on variability for right shoulder flexion and torso rotation (Table 5a). For right shoulder flexion CoV, the GG–corrected test was significant ($p = .007$). Torso rotation CoV also differed by load ($p = .009$). The apparent effect for left shoulder flexion CoV did not remain significant after GG–correction (uncorrected $p = .045$; corrected $p = .056$). All other load main effects were not significant (all $p \geq .187$).

A significant Load × Sex interaction was observed for right hip adduction CoV ($p = .037$); all other interactions were not significant (all $p \geq .097$).

Table 5a. Main effect of Load and Load × Sex interaction on kinematics ROM variability (CoV)

Variable	Baseline	One-hand loading	Two-hand loading	Main effect of Load (p)	Sex × Load (p)
Hip flexion R (CoV, %)	13.48 ± 1.72	14.07 ± 1.43	14.70 ± 1.67	0.546	0.849
Hip flexion L (CoV, %)	14.42 ± 1.18	15.90 ± 1.18	16.26 ± 1.36	0.193	0.366
Knee angle R (CoV, %)	21.17 ± 2.62	18.60 ± 2.02	19.15 ± 2.38	0.455	0.822
Knee angle L (CoV, %)	20.92 ± 2.36	17.67 ± 1.99	18.94 ± 2.65	0.187	0.777
Ankle angle R (CoV, %)	22.67 ± 1.95	24.51 ± 1.91	23.71 ± 1.84	0.313	0.515
Ankle angle L (CoV, %)	19.21 ± 1.40	19.69 ± 1.75	19.51 ± 1.52	0.936	0.438
Hip adduction R (CoV, %)	15.16 ± 1.07	15.55 ± 1.03	14.31 ± 0.97	0.392	**0.037 ***
Hip adduction L (CoV, %)	14.90 ± 1.20	15.47 ± 0.96	16.11 ± 1.36	0.508	0.097 †
Shoulder flexion R (CoV, %)	23.75 ± 2.08	33.51 ± 2.27	29.03 ± 2.64	**0.007* (GG)**	0.506
Shoulder flexion L (CoV, %)	25.26 ± 2.43	31.87 ± 1.67	32.93 ± 3.65	0.045 → (GG = 0.056)	0.697
Torso extension (CoV, %)	17.50 ± 1.25	17.96 ± 0.92	17.82 ± 1.54	0.95	0.945
Torso bending (CoV, %)	16.62 ± 1.26	15.03 ± 0.80	15.98 ± 1.01	0.303	0.274
Torso rotation (CoV, %)	15.27 ± 1.30	16.97 ± 0.89	19.13 ± 1.16	**0.009***	0.197

Note. Values are estimated marginal means ± standard error (SE). †$p < .10$, *$p < .05$, **$p < .01$, ***$p < .001$. For measures with sphericity violations, Greenhouse–Geisser corrected p-values are reported.

The pairwise analysis showed that right shoulder flexion variability was significantly higher during one-hand loading compared with baseline ($p < .001$). In contrast, the differences between baseline and two-hand loading ($p = .262$) and between the two loading conditions ($p = .682$) were not significant. For the left shoulder, only the comparison between baseline and one-hand loading reached significance ($p = .031$). For the torso rotation, variability was greater in the two-hand loading than in baseline ($p = .013$), with no significant differences observed between baseline and one-hand loading ($p = .466$) or between one-hand and two-hand loading ($p = .244$) (Table 5b).

Table 5b. Pairwise load comparisons for kinematics ROM variability (CoV, Bonferroni-adjusted)

Variable	Baseline vs One-hand Δ (95% CI)	p-value	Baseline vs Two-hand Δ (95% CI)	p-value	One-hand vs Two-hand Δ (95% CI)	p-value
Shoulder flexion R (CoV, %)	−9.76 (−15.03, −4.49)	<.001 ***	−5.28 (−12.88, 2.32)	0.262	4.47 (−4.75, 13.70)	0.682
Shoulder flexion L (CoV, %)	−6.61 (−12.74, −0.48)	0.031 *	−7.67 (−17.23, 1.89)	0.152	−1.06 (−9.78, 7.65)	1
Torso rotation (CoV, %)	−1.70 (−4.66, 1.26)	0.466	−3.85 (−7.02, −0.69)	0.013 *	−2.15 (−5.18, 0.88)	0.244

Note. Δ = mean difference (first − second condition) based on estimated marginal means. Positive values mean the first condition is larger. Adjusted p values use Bonferroni. *p < .05. , **p < .01, ***p < .001.

Sex did not show a main effect on kinematic ROM variability (all $p \geq .119$) as shown in Table 5c.

Table 5c. Main effect of Sex on kinematics ROM variability (CoV)

Variable	F(df=1,28)	p-value	Partial η^2	Power
Hip flexion R (CoV, %)	0.255	0.618	0.009	0.078
Hip flexion L (CoV, %)	0.359	0.554	0.013	0.089
Knee angle R (CoV, %)	2.433	0.13	0.08	0.325
Knee angle L (CoV, %)	0.533	0.472	0.019	0.109
Ankle angle R (CoV, %)	0.238	0.63	0.008	0.076
Ankle angle L (CoV, %)	0.475	0.497	0.017	0.102
Hip adduction R (CoV, %)	1.05	0.314	0.036	0.168
Hip adduction L (CoV, %)	0.026	0.873	0.001	0.053
Shoulder flexion R (CoV, %)	0.368	0.549	0.013	0.09
Shoulder flexion L (CoV, %)	0.094	0.761	0.003	0.06
Torso extension (CoV, %)	0.185	0.67	0.007	0.07
Torso bending (CoV, %)	0.079	0.781	0.003	0.058
Torso rotation (CoV, %)	2.589	0.119	0.085	0.343

Note. Asterisks (*) show statistical significance at p < .05.

7. Discussion

In this study, when walking with bilateral hand-held loads, participants showed left side adjustments characterized by shorter stride length, longer stride time, and reduced gait speed. Stride width remained unchanged. For spatiotemporal variability, only the left side stride time CoV increased during two-hand loading, while other variability parameters were unchanged.

Our findings indicate that hand-held loading caused adjustments in spatiotemporal gait regulation, particularly on the non-dominant side (left side). The observed side-specific effect may reflect functional lateralization in healthy gait. Indeed, consistent with previous research, the dominant side tends to focus on propulsion, while the non-dominant limb emphasizes support and ML stabilization. Consequently, the adjustments primarily involve the left (non-dominant) side (Sadeghi et al., 2000; Astephen et al 2007). During treadmill walking, participants modified their stride length, stride time, and walking speed to accommodate the load, consistent with previous research showing that carrying a load generally shortens stride length and increases cadence. (LaFiandra et al., 2003; Liew et al., 2016). Previous research also shows that step width is a crucial control variable for maintaining stability in ML. However, under external loading, the direction of change in step width depends on loading and task constraints. McAndrew and colleagues' treadmill study (2012) demonstrated that, without added external loads and when foot placement was deliberately manipulated, wider steps increased lateral margins of stability, showing a stabilizing benefit of wider steps (McAndrew Young & Dingwell, 2012). In contrast, Baudendistel and colleagues' overground study (2020) with symmetrical bimanual upper-limb loading found that step width decreased and variability increased. The authors interpreted this pattern as reduced adaptability or increased stabilization demand, rather than improved stability (Baudendistel et al., 2020). Therefore, although increased step width usually enhances the lateral margin of stability (McAndrew Young & Dingwell, 2012), in arm-constrained or loaded conditions, participants might use alternative compensatory strategies to maintain stability and require additional control of stability. (Baudendistel et al., 2020; Hausdorff, 2005; Kang & Dingwell, 2008; Sadeghi et al., 2000). In our study, stride width remained unchanged. We believe that the stride width constraint of the treadmill belt may have limited participants' ability to modify lateral foot placement. Instead, it is possible that participants enhanced stabilization by increasing control across the torso, pelvis, and hip, together with unilateral phase adjustments (i.e., small step-to-step foot-contact timing adjustments). These adjustments were expressed primarily in the non-dominant foot. Indeed, the data showed an increase in left side stride time CoV, while the CoV of other spatiotemporal parameters remained essentially unchanged. A higher stride time CoV indicates larger step-to-step fluctuations in temporal rhythm, which reflects reduced gait stability and increased motor control demands (Hausdorff, 2005; Kang & Dingwell, 2008). This effect emerged only in the non-dominant limb, suggesting that it plays a greater compensatory role under load, consistent with previous research showing that the non-dominant limb often exhibits higher variability due to its stabilizing function (Sadeghi et al., 2000).

The shoulder kinematics appeared particularly sensitive to loading. However, differences in shoulder behavior between sides were expected as the unilateral load was applied to the left side. Specifically, right shoulder flexion ROM decreased stepwise from baseline to one-hand to two-hand conditions. In contrast, on the left side, flexion ROM was higher at baseline than under either loading condition. This bilateral reduction is expected when one arm is constrained by a hand-held load. Arm swings are bilaterally coupled, and the control objective is to preserve anti-phase coordination while keeping whole-body angular momentum near zero. Consequently, the contralateral arm also downregulates its swing amplitude, leading to a decrease in shoulder flexion ROM rather than an increase. This pattern aligns with our trunk results (reduced rotation, larger lateral bending) and with biomechanical theories that arm swing is used to minimize trunk rotation torques and stabilize angular momentum (Pontzer et al., 2009; Meyns et al., 2013). Although some studies, under manipulations that permit or encourage contralateral compensation, have reported a pattern in which ROM decreases on the constrained side while increasing on the opposite side (Bondi, 2017; Major et al., 2019), in this study (hand-held load carriage on a treadmill with no instructions to compensate with the free arm) our findings align more closely with reports that arm swings are bilaterally coupled and regulated to minimize whole-body angular momentum. Accordingly, shoulder flexion ROM tends to be downregulated on both sides, accompanied by reduced trunk rotation and other upper-body adjustments (Meyns et al., 2013; Ford et al., 2007; Takami et al., 2020).

Torso ROM showed plane-specific patterns: in the sagittal plane (flexion/extension), ROM was lowest during one-hand loading; in the frontal plane (lateral flexion), ROM was highest during two-hand loading; and in the transverse plane (rotation), ROM decreased under unilateral loading and decreased even more under bilateral loading compared to baseline. These patterns indicate that participants stabilized their gait by reducing shoulder flexion amplitude, decreasing torso rotation, and under two-hand loading, increasing lateral bending. The torso played a compensatory role when arm swing was constrained. Normally, transverse plane trunk rotation works in concert with arm swings to counteract lower-limb angular momentum. When shoulder flexion was reduced by load, participants decreased the amplitude of trunk rotation but increased frontal-plane lateral bending under two-hand loading. This suggests that mechanically, this pattern repositions the center-of-mass projection over the stance foot and likely lowers the hip abductor moment, providing medio-lateral stability without widening the base of support (Perry & Burnfield, 2010; LaFiandra et al., 2003).

For ROM variability, one-hand loading showed higher variability in right shoulder flexion, while two-hand loading increased variability in torso rotation. Increased variability under unilateral loading may reflect inconsistent compensatory strategies of the unloaded arm, as bilateral arm swing is normally coupled and constrains trunk angular momentum (Pontzer et al., 2009; Meyns et al., 2013). When a load restricts one arm, the contralateral arm may attempt to maintain balance with variable amplitude, resulting in greater fluctuations. Under bilateral loading, however, arm swing was restricted on both sides, placing larger demands on the trunk to manage angular momentum. This was expressed as higher variability in torso rotation, consistent with previous studies showing that constraining the arms amplifies trunk kinematic variability (Bruijn et al., 2010; Delabastita et al., 2016). These results suggest that load carriage challenges upper-body coordination stability, with variability concentrated in the body segment responsible for compensation.

Previous studies show that restricting arm swing during walking alters gait performance, reduces stability, and modifies torso kinematics (Bruijn et al., 2010; Delabastita et al., 2016; Ford et al., 2007). Our findings are consistent with this literature. When participants carried loads in one or both hands, arm swing was significantly constrained; shoulder flexion ROM declined, and torso rotation decreased, while variability in trunk kinematics increased (LaFiandra et al., 2003; Liew et al., 2016; Walsh et al., 2018). Under symmetric two-hand loading, we also observed greater torso lateral-flexion ROM, similar to compensatory patterns reported under unilateral loading (Zhang et al., 2010; Matsuo et al., 2008). Taken together, handheld load carriage first constrained natural arm swing (during both unilateral and bilateral loads) and subsequently led to compensatory trunk adjustments. Because stride width did not change, stability was achieved primarily through upper-body reconfiguration, reduced shoulder flexion, decreased torso rotation, and increased lateral bending. This integrated arm–torso strategy redistributed angular momentum control, allowing participants to maintain stable locomotion despite external loading (Perry & Burnfield, 2010; LaFiandra et al., 2003).

Although a significant Sex × Load interaction was found for left hip adduction ROM, post-hoc tests showed that the sex difference was present only at baseline (females $\approx 1.7^{\circ}$ greater than males) and not under either one-hand or two-hand loading. In other words, once a hand-held load was introduced, hip adduction amplitude converged to similar levels across sexes. The mechanism may arise from handheld loading, leading to a similar ML stabilization constraint in both sexes, which increases trunk–pelvis coupling and reduces the available hip degrees of freedom, thereby driving convergence of frontal plane hip amplitude (Meyns et al., 2013; LaFiandra et al., 2003). In contrast, the variability results showed a different pattern: right hip adduction CoV exhibited a main effect of sex ($p = .037$), with males demonstrating larger step-to-step fluctuation across loading conditions. We believe amplitude convergence and variability differences are distinct phenomena. The former indicates that external loading imposes the same movement constraints on both sexes, synchronously reducing the available movement workspace and thereby bringing frontal-plane hip excursion to similar levels. The latter indicates persistent differences in strategy or control style with females regulating frontal-plane hip motion more consistently (lower CoV) and males relying relatively more on trunk or upper-body adjustments, leading to larger hip-level fluctuations (Ferber et al., 2003; McKean

& Burkett, 2012). Overall, loading decreases or eliminates sex differences in amplitude, but sex differences in variability persist; moreover, higher CoV is commonly interpreted as less consistent temporal and kinematic control and larger control demands, and has been linked to reduced gait stability (Hausdorff, 2005; Kang & Dingwell, 2008).

The findings of this study provide clear insight into the extent to which our hypotheses were supported. The first hypothesis stated that hand-held loading would significantly affect spatiotemporal gait parameters and whole-body kinematics compared to unloaded walking. This hypothesis was supported. Participants demonstrated shorter stride length, slower walking speed, longer stride time, and increased stride-time variability under loading conditions. In addition, significant changes in shoulder and trunk ROM were observed, indicating clear upper-body adaptations. These findings confirm that hand-held loading meaningfully modifies gait performance and strategies.

The second hypothesis mentioned that sex would influence gait adaptations under loading. This hypothesis was not supported overall. Although raw stride speed showed significant sex differences, these differences disappeared after height normalization, indicating that the apparent effect was largely attributable to body-size differences rather than sex-specific gait control strategies. Furthermore, while a few isolated sex × load interactions were observed (e.g., left hip adduction ROM), these effects were inconsistent across joints, sides, and loading conditions. Therefore, sex was not considered a primary determinant of gait adaptation in this healthy young sample.

This study has limitations that should be addressed in future work. Firstly, the sample consisted only of young healthy adults. While this design helped control physiological differences and improved internal validity, it also limits the generalizability of our findings to older adults or clinical populations. The effects of hand-held loading on gait may differ across age groups due to changes in balance capacity, gait variability, and muscle strength. Future studies should therefore include older adults and clinical populations to better understand how loading strategies interact with age-related gait adaptations. Furthermore, all walking trials were conducted on a treadmill. Although treadmill walking provides controlled and consistent conditions, it may differ from overground walking in step width regulation, belt-speed constraints, and surface characteristics. These factors could influence gait behavior and reduce ecological validity. Additionally, the loading conditions in this study were limited to 10% BM in unilateral and bilateral dumbbell configurations. This load level is safe and representative of daily hand-held tasks, but real-world loading scenarios are often more complex. Different load magnitudes, carrying methods (e.g., shopping bags, shoulder bags), weight distributions, and additional environmental demands (e.g., stairs, turning) may lead to different gait adaptations. Lastly, although we used a full-body 3D motion-capture system to characterize joint kinematics, we did not collect ground reaction forces or muscle activity. Without force plates and EMG, we were unable to quantify loading-related changes in joint kinetics or muscle activation strategies. Future studies should integrate these tools to provide a more comprehensive analysis of the mechanical and neuromuscular responses to hand-held loading.

In this study, we demonstrated the specific effects of hand-held loading on spatiotemporal gait parameters and whole-body kinematics in healthy young adults. Overall, these findings provide useful insight into how individuals adjust their walking when carrying objects in their hands. Although sex differences were limited in this young sample, previous research from our group suggests that such differences may become more pronounced with aging. Therefore, future studies should extend this work to older adults and clinical populations. Examining responses to symmetric and asymmetric loads, varying load magnitudes, and different carrying methods across treadmill and overground contexts will help determine whether age- or sex-specific compensatory strategies emerge under loading. Ultimately, a better understanding of these adaptations may inform safer loading recommendations, support the development of targeted gait-training approaches, and contribute to maintaining stable and efficient walking in daily life for individuals who experience mobility challenges.

8. Conclusion

This study examined the effects of asymmetric (one-hand loading) and symmetric (two-hand loading) hand-held loading on gait in healthy young adults. Gait stabilization under load was achieved primarily through an integrated arm–torso strategy. With arm swing restricted, trunk rotation decreased. Under two-hand loading, lateral bending increased. Participants maintained ML stability without widening step width. Spatiotemporal adaptations were lateralized to the non-dominant side. Left stride length and speed decreased. Left stride-time variability increased only under two-hand loading, indicating increased compensatory demands. Sex effects were limited. A Sex × Load interaction for left hip adduction ROM appeared only at baseline—females showed more hip adduction than males—and disappeared with loading, suggesting convergence in available hip excursion. In contrast, a main effect of sex persisted for variability, with males exhibiting higher right-hip adduction CoV across conditions. Overall, hand-held loading challenges upper body stabilization more than lateral foot placement. This pattern suggests that stability training and rehabilitation may benefit from targeting arm–torso dynamics. These insights provide a foundation for extending future work to older adults and clinical populations, where hand-held loading is common and gait safety is critical.

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10. Appendices

Appendix A

International Physical Activity Questionnaire (IPAQ)

INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE (August 2002)

SHORT LAST 7 DAYS SELF-ADMINISTERED FORMAT

FOR USE WITH YOUNG AND MIDDLE-AGED ADULTS (15-69 years)

The International Physical Activity Questionnaires (IPAQ) comprises a set of 4 questionnaires. Long (5 activity domains asked independently) and short (4 generic items) versions for use by either telephone or self-administered methods are available. The purpose of the questionnaires is to provide common instruments that can be used to obtain internationally comparable data on health–related physical activity.

Background on IPAQ
The development of an international measure for physical activity commenced in Geneva in 1998 and was followed by extensive reliability and validity testing undertaken across 12 countries (14 sites) during 2000. The final results suggest that these measures have acceptable measurement properties for use in many settings and in different languages, and are suitable for national population-based prevalence studies of participation in physical activity.

Using IPAQ
Use of the IPAQ instruments for monitoring and research purposes is encouraged. It is recommended that no changes be made to the order or wording of the questions as this will affect the psychometric properties of the instruments.

Translation from English and Cultural Adaptation
Translation from English is supported to facilitate worldwide use of IPAQ. Information on the availability of IPAQ in different languages can be obtained at www.ipaq.ki.se. If a new translation is undertaken we highly recommend using the prescribed back translation methods available on the IPAQ website. If possible please consider making your translated version of IPAQ available to others by contributing it to the IPAQ website. Further details on translation and cultural adaptation can be downloaded from the website.

Further Developments of IPAQ
International collaboration on IPAQ is on-going and an ***International Physical Activity Prevalence Study*** is in progress. For further information see the IPAQ website.

More Information
More detailed information on the IPAQ process and the research methods used in the development of IPAQ instruments is available at www.ipaq.ki.se and Booth, M.L. (2000). *Assessment of Physical Activity: An International Perspective*. Research Quarterly for Exercise and Sport, 71 (2): s114-20. Other scientific publications and presentations on the use of IPAQ are summarized on the website.

INTERNATIONAL PHYSICAL ACTIVITY QUESTIONNAIRE

We are interested in finding out about the kinds of physical activities that people do as part of their everyday lives. The questions will ask you about the time you spent being physically active in the **last 7 days**. Please answer each question even if you do not consider yourself to be an active person. Please think about the activities you do at work, as part of your house and yard work, to get from place to place, and in your spare time for recreation, exercise or sport.

Think about all the **vigorous** activities that you did in the **last 7 days**. **Vigorous** physical activities refer to activities that take hard physical effort and make you breathe much harder than normal. Think *only* about those physical activities that you did for at least 10 minutes at a time.

1. During the **last 7 days**, on how many days did you do **vigorous** physical activities like heavy lifting, digging, aerobics, or fast bicycling?

   _____ **days per week**

   ☐ No vigorous physical activities → ***Skip to question 3***

2. How much time did you usually spend doing **vigorous** physical activities on one of those days?

   _____ **hours per day**

   _____ **minutes per day**

   ☐ Don’t know/Not sure

Think about all the **moderate** activities that you did in the **last 7 days**. **Moderate** activities refer to activities that take moderate physical effort and make you breathe somewhat harder than normal. Think only about those physical activities that you did for at least 10 minutes at a time.

3. During the **last 7 days**, on how many days did you do **moderate** physical activities like carrying light loads, bicycling at a regular pace, or doubles tennis? Do not include walking.

   _____ **days per week**

   ☐ No moderate physical activities → ***Skip to question 5***

4. How much time did you usually spend doing **moderate** physical activities on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

Think about the time you spent **walking** in the **last 7 days**. This includes at work and at home, walking to travel from place to place, and any other walking that you have done solely for recreation, sport, exercise, or leisure.

5. During the **last 7 days**, on how many days did you **walk** for at least 10 minutes at a time?

_____ **days per week**

☐ No walking → ***Skip to question 7***

6. How much time did you usually spend **walking** on one of those days?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

The last question is about the time you spent **sitting** on weekdays during the **last 7 days**. Include time spent at work, at home, while doing course work and during leisure time. This may include time spent sitting at a desk, visiting friends, reading, or sitting or lying down to watch television.

7. During the **last 7 days**, how much time did you spend **sitting** on a **week day**?

_____ **hours per day**

_____ **minutes per day**

☐ Don't know/Not sure

This is the end of the questionnaire, thank you for participating.

Appendix B

Consent Form:

uOttawa
Faculté des sciences de la santé
Faculty of Health Sciences

Université d'Ottawa
Faculté des sciences de la santé
École des sciences de l'activité physique

Université d'Ottawa
Faculté de la santé Sciences
École des sciences de l'activité physique

Consent Form

Title of study:

"The effects of (a)symmetric hand-held loads on gait performance in young and older males and females"

Principal Investigator: Jordan Yu
Supervisor: Julie Nantel

This study is part of the PI's academic requirements for the Master's thesis under the supervision of Professor Nantel.

Affiliation: University of Ottawa, Faculty of Health Sciences, School of Human Kinetics

Invitation to Participate: We are inviting you to participate in the abovementioned research study conducted by Jordan Yu and Julie Nantel.

Purpose of the Study: The purpose of the study is to investigate whether symmetric and asymmetric hand loading affects gait performance in young and older males and females.

Specifically, we will examine the effect of three different hand load conditions: (1. Natural walking, no load, 2. One-hand loading at 15% body mass, and 3. Both hands loading at 15% body mass) on gait stability and kinematics. (For example, a 70 kg adult will use 10.5 kg dumbbells as external loads.)

Participation: Your participation will consist of a single session lasting approximately 1.5 hours.

You will first complete the International Physical Activity Questionnaire (IPAQ) (approx. 10 minutes). Then, you will be outfitted with the appropriate motion capture suit and markers (approx. 15 minutes). Next, you will have enough time to warm up and get used to walking on the treadmill and carrying a 15% body mass load. (approx. 15 minutes)

You will walk on a treadmill for 5 minutes under three different loading conditions:

1. Natural walking without loading (baseline),
2. Non-dominant hand loaded with 15% body mass
3. Both hands loaded with 15% body mass.

You will have a 5-10 minute break between each trial (three trials and an optional break totaling approximately 45 minutes).

☎ 613-562-5852
🖶 613-562-5149

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Ottawa ON K1N 6N5 Canada

www.uOttawa.ca

Risks: Your participation in this study will require you to perform three walking trials. Since this study involves walking with hand loads, participants may experience muscle soreness after the data collection session.

We will provide you frequent and adequate breaks to reduce fatigue and muscle soreness caused by walking and carrying loads and assure you that you are allowed at any time to withdraw your participation without adverse consequences.

Benefits: Your participation in this study will help deepen our understanding of the impact of holding symmetrical and asymmetrical loads on gait performance, with a focus on potential variations across different age and sex groups. This knowledge would provide better weight-loading strategies and be critical in developing tailored gait (re)training programs as a function of age and sex.

The parking fee will be reimbursed (parking pass at Lees Campus).

Confidentiality and Privacy: All information shared during the testing session will remain confidential. The contents will be used only for motion analysis and age-matching. Participant's identity will be protected and remain entirely anonymous. Numerical codes and their corresponding data will be assigned to each participant. These codes will be stored separately from the data files on a password-protected computer located in the locked Motor Performance Lab Room 251 at the University of Ottawa Lees Campus.

Conservation of Data: All participant data will be anonymized by a numerical identifier, and the consent form will be stored separately from the data. All electronic data will be stored on a password-protected desktop computer, and a backup of the data will be stored on the external hard drive in the locked Motor Performance lab room 251. Data will be stored on Dr. Nantel's password-protected desktop computer located in the locked Motor Performance Lab Room 251 at the University of Ottawa Lees Campus. All electronic data will be backed up on an external hard located in room 251. Any paper data (i.e. consent forms) will be stored in room 251 in a locked filing cabinet.

As per the guidelines for data collection and retention, the hardcopy data will be retained for 5 years after completion of data collection. Subsequently, all data will be deleted. The retention period will begin when data collection for the study is complete.

Voluntary Participation: You are under no obligation to participate, and if you choose to participate, you can withdraw from the study at any time and/or refuse to answer any questions without suffering any negative consequences. If you choose to withdraw, all data gathered until the time of withdrawal will be removed from the dataset and not used in the study.

If you have any questions about the study, you may contact the researcher or their supervisor. If you have any questions regarding the ethical conduct of this study, you may contact the Office of Research Ethics and Integrity via email (ethics@uottawa.ca) or telephone the Office at 613-562-5800 ext. 5387.

It is recommended that you keep a copy of this consent form for my records.

Acceptance: By signing your name below, you agree to participate in this research study.

Participant's name: ________________________ Date: ___________________

Participant's signature: ________________________ Date: ___________________

Researcher's signature: ________________________ Date: ___________________

Appendix C

To determine whether the observed sex differences in the raw spatiotemporal gait parameters were primarily attributable to anthropometric scaling rather than sex-specific gait control strategies, a secondary height-normalization analysis was conducted. In the original results, left stride length, left stride speed, and toe-based step width showed significant main effects of Sex. Therefore, height normalization was applied only to these three variables to examine whether the sex effect remained after controlling for body size. The results of the normalized analysis are presented in Table C1.

Table C1. Main effect of Sex on height-normalized spatiotemporal gait parameters

Variable	**F(df=1,28)**	**p-value**	**Partial η²**	**Power**
Stride length L (normalized)	1.01	0.328	0.035	0.16
Stride speed L (normalized)	0.99	0.329	0.034	0.16
Stride width Toe (normalized)	2.65	0.116	0.087	0.33

Note. No significant sex effects were observed after normalization (all $p \geq .116$).