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Carrying asymmetric loads while walking on an uneven surface.

Junsig Wang
University of Arizona

Jason C. Gillette
Iowa State University, gillette@iastate.edu

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Abstract

Background Individuals often carry asymmetric loads over challenging surfaces such as uneven or irregular terrain, which may require a higher demand for postural control than walking on an even surface.

Research Question The purpose of this study was to assess postural stability in the medial-lateral (ML) direction while carrying unilateral versus bilateral loads when walking on even versus uneven surfaces.

Methods Nineteen healthy young adults walked on even and uneven surface treadmills under three load conditions: no load, 20% body weight (BW) bilateral load, and 20% BW unilateral load. A Pedar in-shoe pressure system (Novel, Munich, Germany) was used to evaluate center of pressure (COP)-based parameters.

Results Carrying 20% BW bilateral or unilateral loads significantly increased double support ratio. In addition, carrying a 20% BW unilateral load significantly increased coefficient of variation (CV) of double support ratio, CV of ML COP excursion, and CV of ML COP velocity. Walking on an uneven surface significantly increased double support ratio, ML COP excursion, ML COP velocity, and CV of double support ratio. When carrying a 20% BW unilateral load, unloaded limb stance had significantly increased double support ratio and ML COP velocity, although it appears that the loaded limb may be used to make step-by-step adjustments as evidenced by the higher CV of ML COP velocity.

Significance Unilateral load carriage, walking on uneven surfaces, and unloaded leg stance are of particular concern when considering postural stability.

Keywords

Asymmetric load, Gait, Time-to-contact, Uneven surface, Center of pressure

Disciplines

Kinesiology | Kinesiotherapy | Motor Control | Psychology of Movement

Comments

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CARRYING ASYMMETRIC LOADS WHILE WALKING ON AN UNEVEN SURFACE

Junsig Wang[1], and Jason Gillette[2]*
[1]Department of Orthopaedic Surgery, University of Arizona, Tucson, AZ, USA
[2]Department of Kinesiology, Iowa State University, Ames, IA, USA
**Corresponding author. Email address: Jwang0408@email.arizona.edu Tel.: 520-621-2938.*

Highlights

- Carrying a unilateral load resulted in decreased postural stability
- Loaded gait resulted in a more conservative postural strategy than unloaded gait
- Walking on an uneven surface increased ML COP excursion and velocity
- Unloaded limb stance during asymmetric load carriage decreased postural stability

Introduction

Individuals often carry items in one hand instead of both hands during activities of daily living. Asymmetric load carriage is expected to produce a lateral shift of the center of mass and may result in a challenge to postural control during walking [1]. Previous studies have reported that asymmetric load carriage increased medial-lateral (ML) center of pressure (COP) velocities during quiet standing [2] and increased ML COP displacement during gait initiation [3]. However, a recent study reported no differences in ML COP excursion and velocity with bilateral and unilateral shopping bags during quiet standing [4]. The discrepancy in results between previous studies may be due to different load conditions or static versus dynamic conditions and indicates that further study is needed to determine how unilateral load carriage affects postural stability during walking.

Individuals may carry asymmetric loads over challenging surfaces such as uneven or irregular terrain, which may require a higher demand for postural control than an even surface. For example, walking on an irregular surface increased step width and step time variability [5], walking on a loose rock surface increased step width variability and ML center of mass (COM) velocity [6], and walking on a multi-surface terrain increased step variability and ML trunk bending variability [7]. Previous studies have also demonstrated that asymmetric load carriage resulted in significant differences in lower extremity joint moments between loaded and unloaded limbs [1, 8]. These findings support the idea that the uneven surfaces are more challenging for postural control, particularly in the ML direction, and that it is of interest to investigate differences for postural control between loaded versus unloaded limb stance.

Postural stability can be evaluated using time-to-contact (TTC), which is the estimated time it takes the COP to reach the boundary of foot [9]. TTC includes both spatial and temporal

(COP position, velocity and acceleration) aspects of postural control relative to the base of support [10]. TTC has been used to evaluate postural stability and provide a measure of how long an individual has to make a postural adjustment before the COP reaches a two-dimensional boundary of foot [9, 10]. However, this approach has been limited to static tasks such as quiet standing. A standard TTC analysis is challenging to apply to gait since the COP must leave the boundary of one foot and shift to the other foot as the human body progresses forward. Therefore, a modified TTC method was proposed to evaluate postural stability during walking in the current study.

Increasing double support time is believed to be a common strategy to improve postural stability during challenging walking conditions or unstable gait. Several studies have reported that symmetric load carriage with a backpack resulted in increased double support time during walking [11, 12]. In addition, increased double support time may be utilized to improve postural stability for individuals at high risk for falls [13, 14]. Thus, increased double support time is indicative of an attempt to improve postural stability and to avoid loss of balance during challenging walking conditions.

Previous studies have indicated that asymmetric load carriage [2, 3], walking on uneven surfaces [5-7], and unloaded limb stance [1] present postural challenges and/or loading asymmetry in the ML direction. However, we are not aware of any studies that have evaluated postural stability of unilateral load carriage while walking on an uneven surface. Therefore, the purpose of this study was to assess postural stability, particularly in the ML direction, when carrying unilateral versus bilateral loads and when walking on even versus uneven surfaces. We tested the following hypotheses:

1. ML COP velocity would be increased and ML TTC percentage would be decreased during unilateral load carriage as compared to bilateral load carriage.
2. ML COP velocity would be increased and ML TTC percentage would be decreased when walking on an uneven surface as compared to an even surface.
3. ML COP velocity would be increased and ML TTC percentage would be decreased during unloaded limb stance as compared to loaded limb stance for unilateral load carriage.

Methods

Nineteen healthy young adults with an age range of 18 to 30 (14 males and 5 females; age 25.5 ± 3.9 years; height 172.6 ± 5.0 cm; mass 69.7 ± 7.2 kg) participated in this study. Participants were free of any pathology that would affect them while walking on a treadmill or prevent them from being able to carry a 20% body weight (BW) load. Individuals were excluded if they had back, neck, leg, foot, or arm pain. Each participant read and signed an informed consent form approved by the institutional review board of Iowa State University (IRB ID: 16-058). Prior to data collection, potential participants were asked if they had any of the listed areas of body pain as part of a medical history questionnaire, and their ability to carry 20% BW load was confirmed during a warm-up session.

Three load conditions were tested: no load, 20% BW bilateral load, and 20% BW unilateral load (Figure 1). The 20% unilateral load was carried in the participant's dominant hand, while the 20% BW bilateral load was evenly split between both sides of the body (10% BW load carried with each hand). Two hand-held bags were utilized in this study and filled with sealed bags of lead shot to match a load normalized according to each participant's body weight. These normalized loads were based on previous studies that indicated significant kinematic and/or kinetic changes when carrying loads ranging from 10% to 20% BW [1, 11, 12]. Two different treadmills were used for the even and uneven surface conditions (Figure 2). Small wood strips (thickness: 1.27 cm × anterior width: 5 cm × medial-lateral length: 13.5 cm) were used to build the uneven surface treadmill with a random pattern. The participants completed six total conditions (2 surfaces ×3 loads).

The participants were instructed to walk on even and uneven surface treadmills for one minute under the three load conditions as a warm-up session to familiarize with the load carriage conditions [15]. The treadmill velocity was started at 0.22 m/s and then the speed was gradually increased or decreased until the participant signaled that the preferred walking speed had been reached. The six preferred walking speeds were recorded and then the slowest walking speed was selected as a constant walking speed for further data collection. Average walking speeds were 1.1 m/s for no load on the even surface, 0.9 m/s for 20% BW bilateral and unilateral load on the even surface, 0.8 m/s for no load on the uneven surface, and 0.7 m/s for 20% BW bilateral and unilateral load on the uneven surface. Thus, the participants were then asked to walk on the even and uneven surface treadmills (average walking speed 0.7 m/s) for 90 seconds under the three load conditions in a randomized order. Each participant was allowed to rest as much as necessary between conditions, with a minimum break of one minute. A Pedar in-shoe pressure system (Novel, Munich, Germany) was used to collect vertical forces and COP in each foot at 100 Hz. Each in-shoe pressure system consists of 99 sensors that allow the measurement of COP trajectory and pressure distribution (validated up to 60 N/cm^2 with a 3.9% error) [16, 17].

Data processing

To minimize any carryover effect from previous conditions, we analyzed the first 10 strides during the last 30 seconds of each condition. During unilateral load carriage, the loaded limb was on the side of the carried load, while the unloaded limb was the opposite side. Single stance phases and double support phases were determined using the vertical forces, and double support ratio was calculated as a ratio of double support time to single stance time. Heel strike and toe-off were detected with a 5% BW threshold for vertical ground force data [18]. A rectangular base of support for each foot (Figure 3) was defined by the dimensions of the Pedar insole sensor (8.5 cm × 26 cm or 27 cm). The origin of the insole sensor is located at the most posterior and medial point of the sensor, and thus the COP positions were recoded as anterior-posterior and ML coordinates relative to this origin. ML COP excursion and mean ML COP velocity were determined during single stance phases for each foot. ML COP velocities and accelerations were calculated with the first central difference method [19].

ML COP positions, velocities, and accelerations were used to calculate ML TTC using the equation in Figure 3. Since the COP shifts between the boundaries of each foot during walking, the assessment of TTC commonly used during quiet stance was modified. TTC was calculated at each data point and then compared to the remaining single stance time (Figure 4). If the TTC was less than the remaining single stance time, then the TTC value was stored for that time point, indicating a postural adjustment was required during single stance. If the TTC was greater than the remaining single stance time, then the TTC was set to the remaining single stance time, indicating double support begins before a postural adjustment was needed. TTC percentage was then calculated by normalizing TTC by mean remaining single stance. A TTC percentage of 100% indicated that no postural adjustment was required during single stance.

Variability in double support ratio, ML COP excursion, and ML COP velocity was evaluated though coefficient of variation (CV) for ten strides. Increased stride-to-stride variability can be indicative of inconsistent steps and decreased stability during walking [20]. In total, there were seven dependent variables: double support ratio, ML COP excursion, mean ML COP velocity, ML TTC percentage, CV of double support ratio, CV of ML COP excursion, and CV of ML COP velocity. COP-based parameters were calculated using a custom-made Matlab code (Mathworks Inc., Natik, MA).

Statistical analyses

The effects of the different loading conditions and the effects of different surfaces on COP parameters were analyzed using repeated measures Multivariate Analysis of Variance (3×2 MANOVA). To investigate these effects, the higher double support ratio, ML COP excursion, and ML COP velocity were selected between left and right limb stance during each gait cycle. The mean values were then calculated for 10 strides. The lowest ML TTC percentage was determined for 20 steps (left and right legs) in order to focus on the most unstable step. Finally, the higher CVs of COP parameters for 10 strides were selected between left and right limb stance.

Univariate repeated measures Analysis of Variance (ANOVA) was performed when main effects of the MANOVA were significant. Bonferroni post-hoc adjustments were used. To investigate the effect of loaded limb stance vs. unloaded limb stance, the COP parameters were determined for each limb separately, and the Hotelling test was performed. Paired t-tests were performed when a main effect of the Hotelling test was significant. The level of statistical significance for all tests was set at $\alpha < 0.05$. Statistical analyses were performed using SPSS® statistics (version 23; SPSS Inc., Chicago, IL, USA).

Results

<u>Effect of load condition</u>

MANOVA revealed significant main effects of load condition ($p = 0.002$). Univariate ANOVA indicated main effects of load condition on double support ratio, ML COP excursion, ML TTC percentage, CV of double support ratio, CV of ML COP excursion, and CV of ML COP velocity (all $p \leq 0.038$; Table 1). Double support ratio was significantly higher when comparing the 20% BW bilateral load and 20% BW unilateral load to no load (both $p < 0.001$). ML COP excursion was significantly higher when comparing no load to the 20% BW bilateral load ($p = 0.007$). ML TTC percentage was significantly lower when comparing the 20% BW unilateral load to the 20% BW bilateral load ($p = 0.029$). CV of double support ratio, ML COP excursion, and ML COP velocity were significantly higher for the 20% BW unilateral load than no load ($p = 0.004$, $p = 0.050$, $p = 0.003$).

<u>Effect of surface</u>

MANOVA also revealed a significant main effect of surface condition ($p = 0.005$). Univariate ANOVA indicated that double support ratio, ML COP excursion, ML COP velocity, and CV of double support ratio were significantly higher when comparing the uneven surface to the even surface during right limb stance (all $p \leq 0.034$; Table 1). No significant interactions of the load and surface were found.

<u>Effect of loaded limb vs. unloaded limb</u>

The Hoteling test revealed a significant main effect of loaded vs. unloaded limb stance ($p = 0.026$). Univariate paired t-tests indicated that double support ratio and ML COP velocity were significantly higher for the unloaded limb stance as compared to the loaded limb stance ($p = 0.001$, $p = 0.033$; Table 2). However, CV of ML COP velocity was significantly higher for the loaded single limb stance than the unloaded single limb stance ($p = 0.017$; Table 2).

Discussion

Effect of unilateral versus bilateral loads

As expected, ML TTC percentage was decreased when carrying a 20% BW unilateral load as compared to a 20% BW bilateral load. This finding suggested that participants required more rapid postural adjustments during the single stance phase while carrying unilateral loads. However, ML TTC percentage when carrying a unilateral load was not significantly different than no load, indicating that the measure may not be sensitive to postural changes during load carriage. Comparable data are not available for this new approach to TTC, so further tests are needed to determine baseline values.

ML COP excursion and velocity were not changed when comparing the unilateral to bilateral load (Table 1). Conversely, it has been reported that a 20% BW asymmetric load with a single strap bag and a briefcase resulted in increased ML COP excursion and velocity as compared to a symmetric load with a backpack during quiet standing [2]. In addition, it has been reported that a 12% BW asymmetric backpack load resulted in increased ML COP excursion compared to a symmetric backpack load during gait initiation [3]. One potential reason for the difference in findings is that our symmetric loads were carried in the hands and thus were further from the center of mass than a symmetric backpack in the medial-lateral direction. Along this line of thinking, participants would adjust to symmetric loading by constraining their ML COP excursion as compared to unloaded walking. Another potential explanation is that loads carried the hand would result in a lower center of mass than a backpack load, which would result in reduced ML COP excursions for similar upper body postural adjustments.

Participants adjusted to both unilateral and bilateral loads by increasing their double support ratio. Similarly, previous studies have reported increased double support time when carrying a 15% BW load as compared to unloaded walking [11, 12]. However, it has also been reported that a 3 kg asymmetric load did not affect double support time as compared to no load and a bilateral load [4]. This disagreement in findings could be due to the relatively lighter loads in the latter study (15-20% BW versus <5% BW). A greater proportion of double support phase has been seen in high-risk individuals for falls, including older adults [13] and patients with knee osteoarthritis [14]. Greater double support time is a strategy to improve postural stability during unstable gait and may reduce the need for postural adjustments during single stance.

An interesting finding is that stride-to-stride variability in double support ratio, ML COP excursion, and ML COP velocity were significantly higher for the 20% BW unilateral load than for no load (Table 1). Previous studies have suggested that higher stride-to-stride variability in gait parameters reflects inconsistent stepping and decreased postural stability during walking [21, 22]. In addition, CV of temporal-spatial gait parameters have been frequently used to estimate gait variability associated with increased risk of falling [22-25]. In this sense, CV of COP parameters may be a useful estimator of postural stability during load carriage in terms of predicting fall risk. Specifically, an increase in the CVs of double support ratio, ML COP excursion, and ML COP velocity may suggest decreased postural stability during unilateral load carriage.

Effect of even versus uneven surfaces

ML COP velocity was higher for the uneven surface, but there was no difference in ML TTC percentage (Table 1). Similarly, a higher ML COM velocity has been reported with a rocky surface than an even surface, but with no changes in the lateral margin of stability [6]. Thus, COP and COM velocities may be more sensitive to changes than ML TTC percentage, which may be due to changes in ML trunk sway when walking on the uneven surface [7]. In addition, increases in double support ratio, ML COP excursion, and CV of double support ratio were observed with the uneven surface. These results may explained by previous observations of increased step width and step time variability with an uneven surface [5].

Effect of loaded versus unloaded limb

ML COP velocity was higher for unloaded limb stance, but ML TTC percentage was not significantly different (Table 2). Since the carried load is further from the unloaded limb, a higher ML COP velocity may be required to shift body weight to the loaded limb. However, the loaded limb stance showed increased CV of ML COP velocity. As an explanation, it has been suggested that slower walking speeds result in higher stride-to-stride variability [26, 27], which may act to increase local dynamic stability [28]. In addition, double support ratio was increased for the unloaded limb during unilateral load carriage, which is associated with balance challenges [29]. Therefore, these combined results indicated that the unloaded limb stance during unilateral load carriage could be of particular concern for postural control.

There are several limitations to this study. First, COP measures are whole body parameters, so without additional video analysis, we are unable to know where in the body postural adjustments are being made. Postural strategies in upper limb under these load and walking conditions should also be considered. Second, we used treadmills to maintain a consistent, but slow preferred walking velocity (0.7 m/s), when individuals may adjust their preferred and step-to-step velocity in non-lab situations. Third, the TTC methods we used to analyze gait are new, so we don’t have previous values to judge ‘good’ and ‘bad’ values for postural stability.

This study is the first that we know of to investigate the effect of asymmetric loads while walking on an uneven surface. Unilateral loads, walking on uneven surfaces, and the unloaded leg are of concern when considering postural stability during load carriage. Our modified time-to-contact method may be of value for future studies that further evaluate postural stability in a balance-challenged population, such as the elderly.

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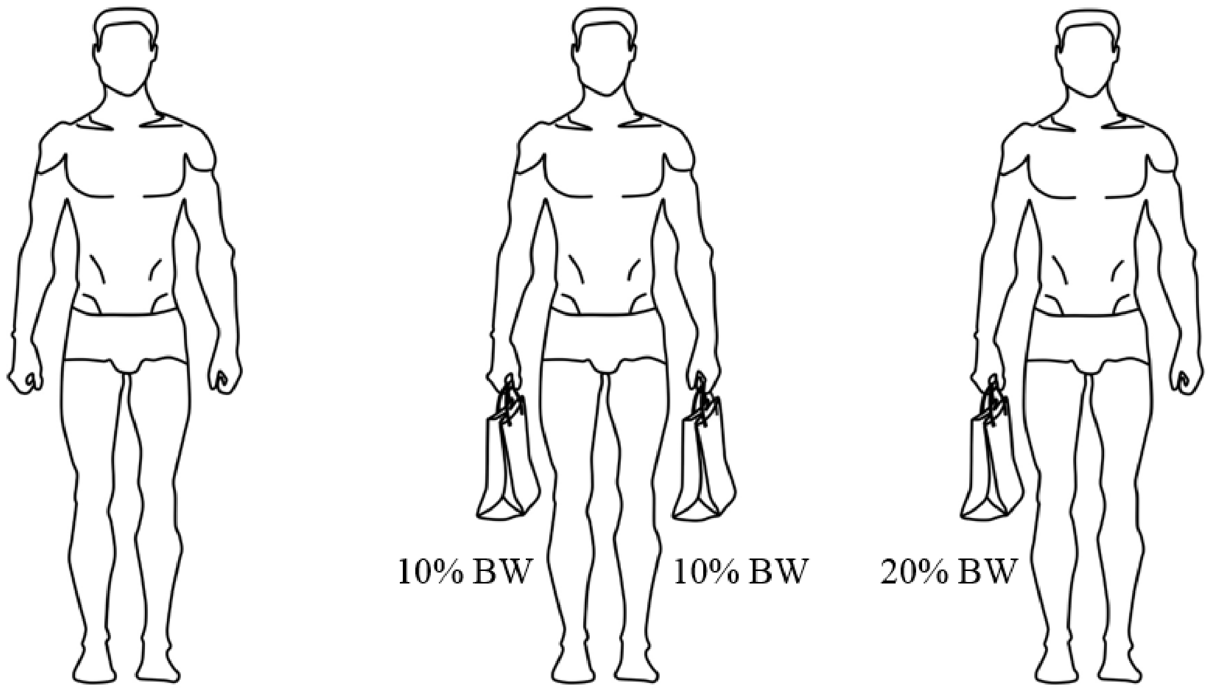


Figure 1

(a) (b)

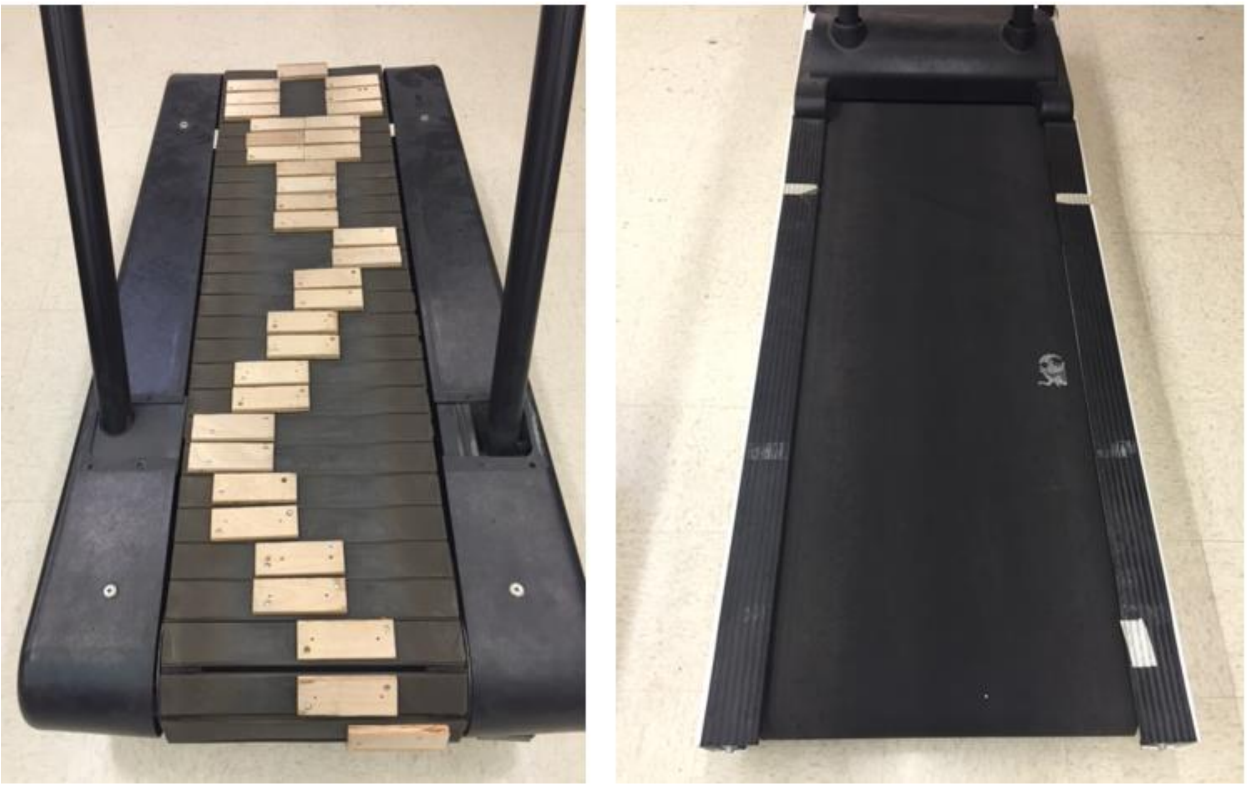
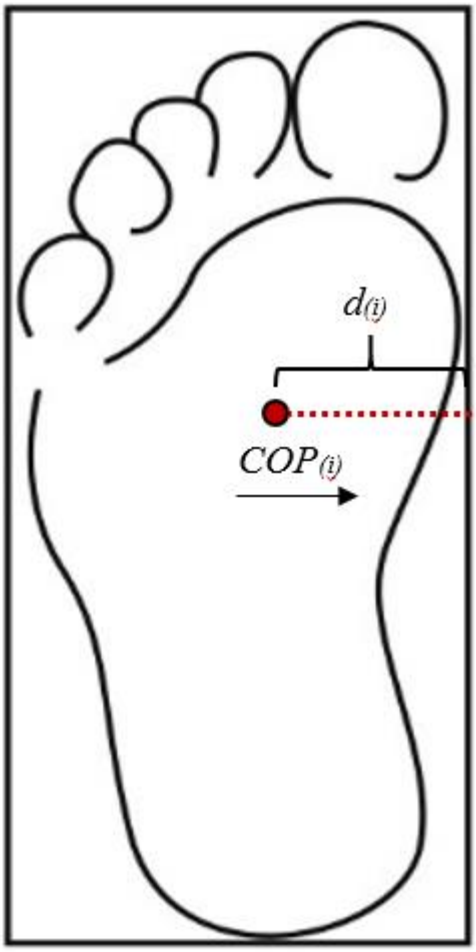


Figure 2



$$TTC_{(i)} = \frac{-v_{(i)} \pm \sqrt{v^2{}_{(i)} + 2a_{(i)}d_{(i)}}}{a_{(i)}}$$

$d_{(i)}$: distance from ML boundary to COP

$v_{(i)}$: ML COP velocity$a_{(i)}$: ML COP acceleration

i : each data point (100 Hz)

Figure 3

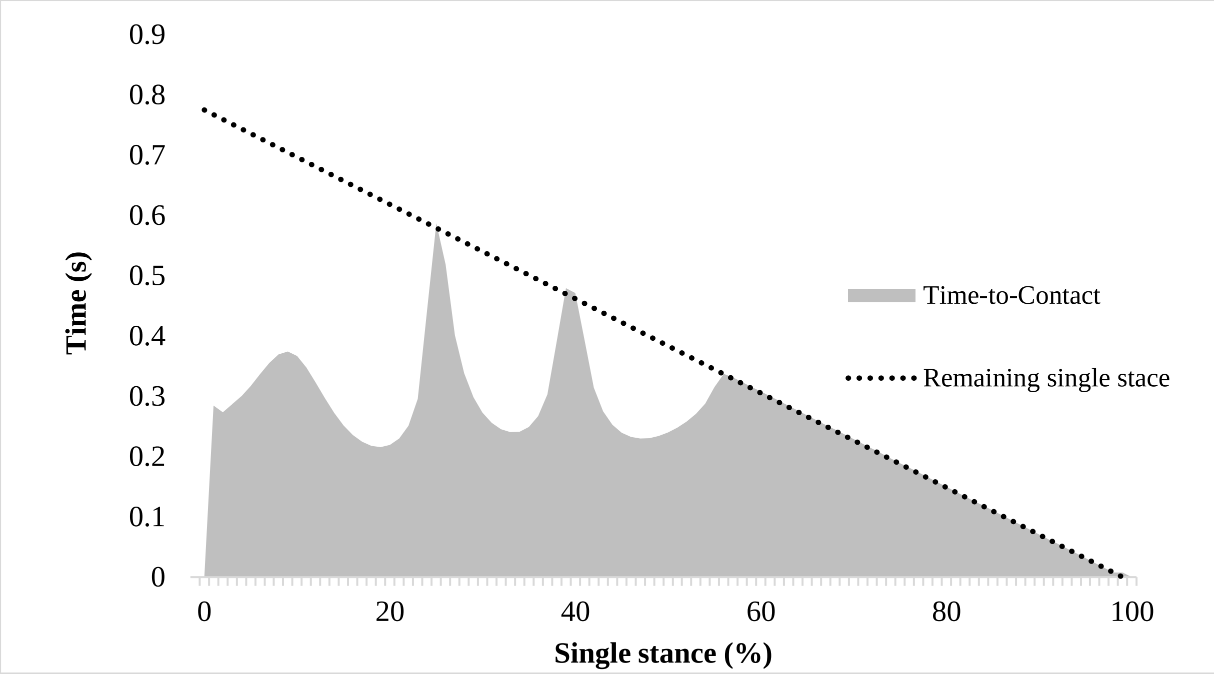

0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
Time (s)
0
20
40
60
80
100
Single stance (%)
Time-to-Contact
Remaining single stace

Figure 4

Captions

Figure 1. Illustration of the three load conditions: no load (left), 20% BW bilateral load split between both sides of the body (center), and 20% body weight (BW) load on one side of the body (right).

Figure 2. Two walking surfaces: (a) uneven surface treadmill vs. (b) even surface treadmill.

Figure 3. Illustration of the rectangular boundary of the foot and the Time-to-Contact (TTC) calculation. v and a (COP velocity and acceleration) were calculated using the first central difference method.

Figure 4. Illustration of TTC during walking: if TTC is less than remaining single stance time, then TTC saved (white appears below the remaining single stance line, single stance adjustment needed). If TTC is greater than remaining single stance, then TTC is equal to remaining single stance time (no white appears below single stance line, no adjustment needed). TTC percentage was calculated by the ratio of TTC and remaining single stance (100% if no adjustment needed).

Table 1. Mean and standard deviations for postural stability parameters under the three load conditions for the even surface and uneven surface.

	No load Mean (SD)	Bilateral Mean (SD)	Unilateral Mean (SD)	Even Mean (SD)	Uneven Mean (SD)	Load p-value	Surface p-value	Load × Surf p-value
Double support ratio	0.471 (0.034)	**0.520[a] (0.036)**	**0.518[a] (0.040)**	0.496 (0.044)	**0.510[*] (0.042)**	**$F_{(2,36)} = 77.50$ $p < 0.001$ $\eta^2 = 0.551$**	**$F_{(1,18)} = 6.242$ $p = 0.022$ $\eta^2 = 0.056$**	$F_{(2,36)} = 0.792$ $p = 0.452$ $\eta^2 = 0.000$
ML COP excursion (mm)	**14.12[b] (3.89)**	12.44 (4.297)	12.89 (4.10)	12.65 (4.36)	**13.65[*] (3.85)**	**$F_{(2,36)} = 6.710$ $p = 0.005$ $\eta^2 = 0.129$**	**$F_{(1,18)} = 5.286$ $p = 0.034$ $\eta^2 = 0.064$**	$F_{(2,36)} = 0.084$ $p = 0.894$ $\eta^2 = 0.001$
ML COP velocity (mm/s)	31.1 (9.0)	30.8 (11.0)	32.4 (11.2)	29.4 (10.3)	**33.4[*] (10.2)**	$F_{(2,36)} = 1.222$ $p = 0.306$ $\eta^2 = 0.023$	**$F_{(1,18)} = 18.37$ $p < 0.001$ $\eta^2 = 0.177$**	$F_{(2,36)} = 0.447$ $p = 0.590$ $\eta^2 = 0.007$
Minimum ML TTC percentage	93.3 (6.0)	95.3 (4.7)	**92.9[b] (5.4)**	94.7 (5.5)	93.0 (5.3)	**$F_{(2,36)} = 4.115$ $p = 0.028$ $\eta^2 = 0.073$**	$F_{(1,18)} = 3.884$ $p = 0.064$ $\eta^2 = 0.043$	$F_{(2,36)} = 0.402$ $p = 0.667$ $\eta^2 = 0.008$
CV double support ratio	0.049 (0.022)	0.056 (0.028)	**0.063[a] (0.024)**	0.049 (0.023)	**0.063[*] (0.026)**	**$F_{(2,36)} = 9.016$ $p = 0.001$ $\eta^2 = 0.111$**	**$F_{(1,18)} = 21.87$ $p < 0.001$ $\eta^2 = 0.185$**	$F_{(2,36)} = 0.319$ $p = 0.665$ $\eta^2 = 0.000$
CV ML COP excursion	0.376 (0.126)	0.394 (0.144)	**0.439[a] (0.113)**	0.404 (0.141)	0.402 (0.119)	**$F_{(2,36)} = 3.587$ $p = 0.038$ $\eta^2 = 0.076$**	$F_{(1,18)} = 0.009$ $p = 0.923$ $\eta^2 = 0.000$	$F_{(2,36)} = 0.362$ $p = 0.696$ $\eta^2 = 0.007$
CV ML COP velocity	0.318 (0.083)	0.355 (0.124)	**0.377[a] (0.084)**	0.350 (0.111)	0.350 (0.091)	**$F_{(2,36)} = 6.063$ $p = 0.006$ $\eta^2 = 0.108$**	$F_{(1,18)} = 0.001$ $p = 0.975$ $\eta^2 = 0.000$	$F_{(2,36)} = 0.290$ $p = 0.726$ $\eta^2 = 0.005$

[a] $p < 0.05$ vs. no load, [b] $p < 0.05$ vs. 20% BW bilateral load, [*] $p < 0.05$ even surface vs. uneven surface

Effect sizes were computed with the eta-squared (η^2) and η^2 = 0.01, 0.06, and 0.14 correspond to small, medium, and large effects.

Table 2. Mean and standard deviations for postural stability parameters for loaded limb stance vs. unloaded limb stance

	Loaded limb Mean (SD)	Unloaded limb Mean (SD)	p-value
Double support ratio	0.489 (0.034)	**0.509*** **(0.042)**	**$t_{(18)} = 3.746$** **$p = 0.001$** **$d = 0.859$**
ML COP excursion (mm)	10.09 (3.72)	11.01 (4.15)	$t_{(18)} = 1.505$ $p = 0.149$ $d = 0.528$
ML COP velocity (mm/s)	25.4 (9.6)	**28.5*** **(11.3)**	**$t_{(18)} = 2.300$** **$p = 0.033$** **$d = 0.528$**
Minimum ML TTC percentage	94.5 (4.5)	94.9 (4.0)	$t_{(18)} = 0.165$ $p = 0.714$ $d = 0.038$
CV double support ratio	0.053 (0.013)	0.055 (0.024)	$t_{(18)} = 0.425$ $p = 0.676$ $d = 0.098$
CV ML COP excursion	0.396 (0.086)	0.347 (0.104)	$t_{(18)} = -1.836$ $p = 0.082$ $d = 0.421$
CV ML COP velocity	**0.348*** **(0.064)**	0.296 (0.086)	**$t_{(18)} = -2.615$** **$p = 0.017$** **$d = 0.600$**

* $p < 0.05$; Effect sizes were computed with Cohen’s D and d = 0.2, 0.5, and 0.8 correspond to small, medium, and large effects.