

## **Mediolateral postural stability when carrying asymmetric loads during stair negotiation**

*Junsig Wang<sup>1,\*</sup> and Jason Gillette<sup>2</sup>*

<sup>1</sup>*Department of Orthopaedic Surgery, University of Arkansas for Medical Science, Little Rock, AR, USA*

<sup>2</sup>*Department of Kinesiology, Iowa State University, Ames, IA, USA*

*\*Corresponding author. Email address: [JWang3@uams.edu](mailto:JWang3@uams.edu) Tel.: 501-246-4439.*

## Abstract

The purpose of this study was to assess postural stability in the medial-lateral (ML) direction when carrying unilateral and bilateral loads during stair negotiation. Twenty-four healthy young adults were instructed to ascend and descend a three step staircase under three load conditions: no load, 20% body mass (BM) bilateral load, and 20% BM unilateral load. A modified time-to-contact (TTC) method was proposed to evaluate postural stability during stair negotiation. Carrying unilateral loads required more rapid postural adjustments as evidenced by lower minimum ML TTC and ML TTC percentage as compared bilateral loads and no load during stair descent. In addition, lower ML TTC and TTC percentage were found for loaded limb stance for stair descent. Taken together, unilateral loads and the loaded leg during stair descent are of concern when considering postural stability during load carriage. These results illustrate differing postural control challenges for stair ascent and descent during load carriage.

**Keywords:** asymmetric loads, postural control, stair negotiations, time-to-contact, center of pressure

## Introduction

Individuals often carry items in one hand during activities of daily living, which results in frontal plane load asymmetry. Carrying loads may improve postural stability by lowering the center of mass (COM) (Rosker et al., 2011). However, load asymmetry may require postural changes such as lateral trunk bending due to a shift of the COM (DeVita et al., 1991) and lead to difficulties in maintaining postural control during walking. Previous studies have reported that an asymmetric load resulted in increased mediolateral (ML) center of pressure (COP) velocity during quiet standing (Anker et al., 2008; Zultowski and Aruin, 2008) and higher ML COP displacement during gait initiation (Fraga et al., 2016) as compared to bilateral loads. More recently, a study has found that asymmetric load carriage with a hand-held bag required more rapid ML postural adjustments as compared to bilateral load carriage during gait (Wang and Gillette, 2018b). However, two studies did not find differences in ML COP parameters between asymmetric and bilateral load carriage during gait (Bampouras and Dewhurst, 2016; Hill et al., 2018). Therefore, there is conflicting evidence whether or not asymmetric load carriage affects postural control during gait, particularly in the ML direction. Moreover, these studies have been limited to lifting or level walking tasks. Thus, further studies on load carriage are needed to assess postural stability for challenging walking conditions such as stair negotiation.

Stair ascent and descent may require higher demand on postural control than level walking due to kinematic, kinetic, and muscle activity differences. When ascending or descending stairs, one limb supports the body mass while the other limb advances to the next step. Krebs et al. (1992) found greater trunk range of motion in the frontal plane for stair ascent and descent than walking. Madehkhaksar and Egges (2016) reported greater ML COM range of motion for stair descent than walking. Other studies have indicated that stair negotiation requires

greater frontal plane knee and hip range of motion and moments, along with altered lower limb muscle coordination for ML postural control compared to walking (Costigan et al., 2002; Lin et al., 2015; Nadeau et al., 2003). These differences likely result in higher demands on postural control for stair negotiation than level walking. Indeed, a study has found that a faster step rate for stair descent was associated with more prospective falls over 12 months and higher fall risk than level walking (Wang et al., 2017). Therefore, it is important to investigate ML postural control during stair ascent and descent for safety considerations.

Previous studies have indicated that asymmetric load carriage resulted in significant differences in lower extremity joint moments between loaded and unloaded limbs during walking (DeVita et al., 1991; Matsuo et al., 2008) and stair negotiation (Wang and Gillette, 2018a). These differences are likely due to a larger moment arm from the load to the stance leg on the opposite side of the body. Another study found that weight-bearing asymmetry in lower limbs resulted in different postural control strategies between the loaded limb and unloaded limb during perturbed standing (de Kam et al., 2016). With these differences between limbs, it is of interest to analyze if there are differences in postural control between the loaded and unloaded limbs.

Falls involving stairs are common accidents resulting in tremendous cost burden to workers. Fall accidents are considered a major source of occupational injuries accounting for US \$15.36 billion or 27.7% of total injury costs (Liberty Mutual Research Institute for Safety, 2019). In addition, fall accidents on the stairs while carrying loads likely result in more dangerous accident scenarios. However, postural control while carrying loads during stair negotiation remains largely unknown. Thus, the purpose of this study was to investigate postural control when carrying symmetric vs. asymmetric loads during stair negotiation. Postural stability in the ML direction was evaluated consistent with previous studies investigating the effect of

asymmetric loads (Anker et al., 2008; Fraga et al., 2016; Zultowski and Aruin, 2008).

Furthermore, it has been suggested that ML COP parameters are linked to higher risk of falls (Maki et al., 1994). Time to contact (TTC) appeared to be effective to evaluate postural stability when carrying asymmetric and bilateral loads (Wang and Gillette, 2018b). In fact, the ML TTC measures were decreased while other COP parameters were not changed when comparing to asymmetric load carriage to symmetric load carriage. Therefore, we hypothesize that: 1) minimum ML TTC and TTC percentage would be decreased for 20% body mass (BM) unilateral loads as compared to 20% BM bilateral loads, and 2) minimum ML TTC and TTC percentage would be decreased for loaded limb stance as compared to unloaded limb stance when carrying a 20% BM asymmetric load.

## **Methods**

Twenty-four healthy young adults (11 males and 13 females; age  $22.1 \pm 2.9$  years; height  $172.8 \pm 8.9$  cm; mass  $73.3 \pm 12.8$  kg) participated in this study. G\*Power was used to calculate a sample size of 20 using previously published data to determine a minimum estimated effect size of 0.67 with an alpha error probability of 0.05 and a power of 0.80. Participants were free of any pathology that would affect them while walking on stairs or prevent them from being able to carry a 20% BM load. Each participant provided informed consent as approved by the university's institutional review board prior to the study.

Three load conditions were tested: no load, 20% BM bilateral load, and 20% BM unilateral load (Figure 1). The load was evenly split between the right and left hands during the bilateral load condition (10% BM in each hand). Two hand-held bags (width: 30cm  $\times$  height: 13cm  $\times$  depth: 25cm) were filled with sealed bags of lead shot to match the loaded conditions. The unilateral load was carried in the participant's dominant hand. The load carried in the bags

was normalized according to each participant's body mass. The level of normalized load was based on the upper range of previous studies that indicated significant kinematic and/or kinetic changes when carrying loads ranging from 10% to 20% BM (Chow et al., 2005; DeVita et al., 1991; Fowler et al., 2006; Hong and Li, 2005).

Participants were instructed to ascend and descend a three-step staircase (step height 18.5 cm, tread depth 29.5 cm) at preferred pace using an unloaded leg lead and a loaded leg lead for each load condition. The order of the conditions was randomized through a list randomizer giving six random numbers (3 load conditions  $\times$  2 leg leads) (Haahr, 1998), and each condition was repeated three times for a total of 36 trials (3 load conditions  $\times$  ascent vs. descent  $\times$  unloaded vs. loaded leg leads  $\times$  3 trials). Both an unloaded and a loaded leg lead were tested to avoid results being biased by any differences that might occur when comparing step 1 vs. step 2 of stair negotiation.

Eight cameras (Vicon, Oxford, UK) were used to collect three-dimensional kinematic data. Twelve markers were placed on both feet: bilateral great toe, heel, medial midfoot, lateral midfoot, medial malleolus, and lateral malleolus. During walking trials (stair ascent and descent), bilateral medial malleolus and medial midfoot markers were removed and recreated using surrounding markers. Two force platforms (AMTI, Watertown, MA) placed on the first and second steps to collect COP data. Video data were sampled at 160 Hz, and force platform data were sampled at 1600 Hz. Video and force platform data were synchronized using Vicon Nexus (Vicon, Oxford, UK). Both video and force platform data were filtered with a fourth-order, low-pass Butterworth filter at a cutoff frequency of 6 Hz. The force data were downsampled so that video and force data both had corresponding data points.

### **Data processing**

During unilateral load carriage, the loaded limb was on the side of the carried load. Single stance and double stance were determined using the vertical ground reaction forces (bottom two steps and ground) and toe velocities (top step). Toe velocities were used to define the gait events for the top staircase step because of the limited number of force plates. Double stance ratio was calculated as a ratio of double stance time to single stance time. ML COP excursion and mean ML COP velocity were determined during single stance phases for each foot. ML COP velocities and accelerations were calculated utilizing the first central difference method. Rectangular boundaries for each foot (Figure 2) were determined using the heel, toe, medial midfoot, and lateral midfoot markers.

ML COP positions, velocities, and accelerations were used to calculate ML TTC using the equation in Figure 2. Since the COP shifts between the boundaries of each foot during walking, a modified version of TTC was utilized (Wang and Gillette, 2018b). In brief, TTC was calculated at each data point and then compared to the remaining single stance time (Figure 3). 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. Minimum ML TTC during single stance was determined as depicted in Figure 3, indicating the fastest postural adjustment during single stance phase. TTC percentage was then calculated by dividing average TTC during single stance by one-half of single stance time. Thus, a TTC percentage of 100% indicated that no postural adjustments were required during single stance.

Numerous studies have used TTC to evaluate postural stability (DiDomenico et al., 2015; Haddad et al., 2006; Hertel and Olmsted-Kramer, 2007; Slobounov et al., 1998; van Wegen et al., 2002). TTC is the estimated time it takes the COP to reach the boundary of foot (Slobounov et al., 1998). TTC includes both spatial and temporal (velocity and acceleration) aspects of

postural control relative to the base of support (Haddad et al., 2006). It has been suggested that TTC is more sensitive and effective than traditional COP parameters in studying postural control in various balance-challenged populations. For example, TTC has been used to detect postural control deficits in the elderly (van Wegen et al., 2002), individuals with ankle instability (Hertel and Olmsted-Kramer, 2007), individuals with knee pain (Rodrigues et al., 2014), and patients with a mild traumatic brain injury (Slobounov et al., 2008), while traditional COP measures did not detect major changes. Another study has also shown that TTC measures of the COP and COM were at least as reliable as the traditional COP parameters (Wheat et al., 2012). In addition, a more recent study has indicated that a simple modification of the TTC was feasible to evaluate gait stability when carrying unilateral vs. bilateral loads (Wang and Gillette, 2018b). However, this approach has mostly been limited to static tasks such as quiet standing instead of dynamic motions. 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 human body progresses forward. Therefore, a modified TTC method was used to evaluate postural stability during stair negotiation in the current study (Wang and Gillette, 2018b).

### **Statistical analyses**

There were five dependent variables: double stance ratio, ML COP excursion, mean ML COP velocity, minimum ML TTC, and ML TTC percentage. COP parameters were calculated using custom-made Matlab code (Mathworks Inc., Natick, MA). The effect of loading condition (Load) was analyzed using repeated measures Multivariate Analyses of Variance (MANOVA). To investigate these effects, the lower ML TTC percentage was selected between step 1 and step 2 in order to focus on the more unstable step. The remaining COP parameters were also determined during the step when the ML TTC percentage is at its minimum in order to

understand a more complete picture of postural control during the more unstable step. Mean values of three trials were used for statistical analyses. Individual repeated measures Analyses of Variance (ANOVA) were performed when main effects were significant. Bonferroni post-hoc adjustments were applied. To investigate the effect of loaded limb stance vs. unloaded limb stance (Limb), the COP parameters were determined for each limb separately, and the Hotelling test was performed. The “loaded” and “unloaded” labels refer to sides which are loaded during the unilateral load condition. These labels were applicable only when carrying 20% unilateral loads and thus the separate statistical analysis was performed focusing on the effect of Limb. Univariate 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**

### Effect of Limb (for a 20% BM asymmetric load)

During stair ascent, the Hotelling test revealed a significant main effect of stance limb ( $p = 0.001$ ). Paired t-tests indicated that ML COP excursion was higher for the loaded limb stance than the unloaded limb stance (Table 1). During stair descent, the Hotelling test also revealed a significant main effect of stance limb ( $p = 0.004$ ). Paired t-tests indicated that ML COP velocity was higher, while minimum ML TTC and ML TTC percentage were lower for loaded limb stance than unloaded limb stance (Table 1). However, double stance ratio was increased for unloaded limb stance as compared to loaded limb stance.

### Effect of Load

#### *Ascent*

MANOVA revealed a significant main effect of load condition ( $p < 0.001$ ). Univariate ANOVA indicated significant main effects of load condition on double stance ratio, ML COP excursion, mean ML COP velocity, ML TTC percentage, and minimum ML TTC (all  $p < 0.001$ ; Table 2). Double stance ratio, ML COP excursion, and mean ML COP velocity were significantly higher for the 20% BM bilateral and unilateral loads than no load (all  $p \leq 0.009$ ). In addition, ML TTC percentage and minimum ML TTC were significantly lower when comparing the 20% BM bilateral and unilateral loads to no load (all  $p \leq 0.008$ ).

### *Descent*

MANOVA revealed a significant main effect of load condition ( $p = 0.016$ ). Univariate ANOVA showed significant main effects of the load condition on double stance ratio, ML COP excursion, ML COP velocity, ML TTC percentage, and minimum ML TTC (all  $p \leq 0.022$ ; Table 2). Double stance ratio was significantly higher for the 20% BM bilateral and unilateral loads than no load ( $p = 0.034$ ,  $p = 0.007$ ). ML COP excursion and velocity were significantly higher for the 20% BM unilateral load than no load ( $p = 0.027$ ,  $p = 0.001$ ). ML TTC percentage and minimum ML TTC were significantly decreased when comparing the 20% BM unilateral load to the 20% BM bilateral load ( $p = 0.044$ ,  $p = 0.004$ ). In addition, minimum ML TTC was significantly lower for the 20% BM unilateral load than no load ( $p = 0.018$ ).

### **Discussion**

The purpose of this study was to investigate the effect of loaded vs. unloaded stance limb and the effect of load asymmetry on postural control during stair ascent and stair descent.

#### Effect of limb (for a 20% BM asymmetrical load)

The effect of loaded limb stance vs. unloaded limb stance was tested when carrying a 20% BM unilateral load. When comparing stance limbs, changes in several stability parameters

were found during stair descent, while most stability parameters remained similar during ascent. For example, loaded limb stance required more ML postural adjustment and faster ML COP velocity when carrying asymmetric loads during stair descent. These results may be due to a greater downward momentum resulting from the additional load for the loaded limb while descending. Furthermore, since the load carried was further from the unloaded limb, faster postural adjustments in the ML direction may be required to shift the center of mass to the loaded limb. These findings suggest that lowering the increased body mass during the loaded limb stance might be a greater challenge in stabilizing body posture than lowering the unloaded limb.

### Effect of Load

#### *Stair Ascent*

Load asymmetry did not affect ML postural stability during stair ascent. Our results suggest that carrying loads is more challenging for postural control than unloaded stair ascent and requires faster ML postural adjustments during single stance phase as evidenced by decreased minimum ML TTC and ML TTC percentage. Previous studies reported decreased ML dynamic stability determined by maximum Lyapunov exponents when comparing a 20 kg (about 29% BM load) backpack (Qu, 2013) and a 12.7 kg (about 16% BM load) vest (Liu and Lockhart, 2013) to no load. Although these studies did include a comparison between unilateral loads and no load, the findings imply that carrying additional anterior/posterior loads decrease ML postural stability during gait. This might be due to increased co-activation of the trunk muscles, which results in higher spine stiffness (Vera-Garcia et al., 2006). Therefore, 20% BM loads likely require a higher demand for ML postural control regardless of ML load asymmetry during stair ascent.

Other COP parameters (ML COP excursion and ML COP velocity) were also not changed when comparing bilateral to unilateral loads during stair ascent. However, Zultowski and Aruin (2008) found increased ML COP excursion and velocity with unilateral single strap bag and briefcase loads than bilateral backpack loads during quiet standing. In addition, Fraga et al. (2016) found increased ML COP excursion with a unilateral than a bilateral backpack load during gait initiation. The difference in results for the current study may be due to static vs. dynamic movement conditions. For double stance ratio, Hong and Li (2005) reported no differences in double stance time between unilateral and bilateral loads with a backpack and strap bag during stair ascent and descent. More recently, Bampouras and Dewhurst (2016) also found that a 3 kg unilateral shopping bag did not affect double stance time as compared to a bilateral shopping bag. Our results also support that load asymmetry does not change traditional COP parameters and double stance time.

In comparison with no load, participants adjusted to both unilateral and bilateral loads by increasing their double support time in response to increased ML COP excursion and ML COP velocity. Conversely, Wang and Gillette (2018b) reported that unilateral and bilateral loads decreased ML COP excursion and decreased ML COP velocity during treadmill walking, which is inconsistent with our results. The difference in findings may be due to differences between level walking and stair negotiation, which indicate that carrying loads during stair ascent may require greater demand on ML postural control. In addition, previous studies found that both bilateral and unilateral loads increased double support time during gait (Crowe et al., 1993; Park et al., 2018). A greater proportion of double support phase is considered a strategy to avoid losing balance in high fall-risk individuals, including older adults (Maki, 1997) and patients with knee osteoarthritis (Gök et al., 2002). Furthermore, lifting of the COM from step to step with the

additional lateral loads may be a balance-challenging task, and thus increased double stance time may be a necessary strategy to maintain dynamic balance and reduce the likelihood of rapid postural adjustments during single stance.

### *Stair descent*

Across parameters, only TTC measures were changed in response to unilateral loads and thus these parameters may be more sensitive to postural changes during asymmetric load carriage. A recent study also suggested that a modified TTC method was feasible to detect postural changes between unilateral to bilateral load carriage during gait (Wang and Gillette, 2018b). In the literature, it has been shown that a TTC approach was more effective in detecting postural changes than traditional COP measures. Thus, the TTC measures may be of value to identify hazardous walking conditions and tasks in order to reduce fall-related injuries.

Interestingly, increased double stance ratio was not changed when comparing the unilateral to bilateral loads. This finding indicates that individuals need similar time constraints of single stance phase when carrying both unilateral and bilateral loads, but they likely required to make more rapid postural adjustments (evidenced by lower TTC values) when carrying the unilateral loads.

Changes in ML COP excursion and velocity were found when comparing the unilateral load to no load. However, Wang and Gillette (2018b) reported decreased ML COP excursion and unchanged ML COP velocity when comparing the unilateral and bilateral loads to no load during gait. The discrepancy between studies may again be due to differences between level walking and stair negotiation. Stair descent may demand greater balance control when stepping down and lowering the COM than level walking. It was suggested that individuals tend to increase ML COP range when stabilizing their balance during challenging walking conditions such as a

slippery surface (Oates et al., 2005). In addition, faster COP movement may be associated with stair descent due to increased downward momentum being transferred to lateral momentum during the braking phase, which may be indicative of greater braking force (Oates et al., 2005). Accordingly, greater ML COP excursion and velocity might be needed when descending stairs with unilateral loads and thus may indicate potentially increased postural risk.

### Limitations and summary

There are several limitations of this study. First, a three step staircase was used, so the participants may not have achieved a repeatable postural control pattern. However, the results may indicate important postural adjustments are required for the first two steps of stair ascent and descent. Second, COP measures reflect whole body postural control, so without further video analysis, we are unable to know where in the body adjustments are being made. Third, we only focused on 20% BM loads, so it is not clear if lower loads such as 10% or 15% BM would have also resulted in the same changes. Fourth, the modified TTC methods used in the current study have only been used in a few studies and thus additional studies are needed to test that these parameters are a valid indicator of postural challenge.

In summary, COP based parameters were changed when comparing both unilateral and bilateral loads to no load during stair ascent, indicating postural adjustments for load carriage in general rather than asymmetry. For stair descent, carrying unilateral loads required more rapid postural adjustments in response to asymmetrical loading as evidenced by lower minimum ML TTC and ML TTC percentage as compared to bilateral loads. In addition, lower ML TTC and TTC percentage were found for loaded limb stance for stair descent. Taken together, unilateral loads and the loaded leg during stair descent are of particular concern when considering postural stability during load carriage. Our results also illustrate differing postural control challenges for

stair ascent and descent during load carriage. Therefore, load carriage recommendations include carrying less than 20% BM loads and splitting loads bilaterally in order to avoid risk of postural instability and fall-related injury. Furthermore, the modified time-to-contact method may be of value for future studies that evaluate postural stability in hazardous walking conditions and load carriage tasks.

## References

- Anker, L.C., Weerdesteyn, V., van Nes, I.J.W., Nienhuis, B., Straatman, H., Geurts, A.C.H., 2008. The relation between postural stability and weight distribution in healthy subjects. *Gait Posture* 27, 471-477. <https://doi.org/10.1016/j.gaitpost.2007.06.002>
- Bampouras, T.M., Dewhurst, S., 2016. Carrying shopping bags does not alter static postural stability and gait parameters in healthy older females. *Gait Posture* 46, 81-85. <https://doi.org/10.1016/j.gaitpost.2016.02.017>
- Chow, D.H.K., Kwok, M.L.Y., Au-Yang, A.C.K., Holmes, A.D., Cheng, J.C.Y., Yao, F.Y.D., Wong, M.S., 2005. The effect of backpack load on the gait of normal adolescent girls. *Ergonomics* 48, 642-56. <https://doi.org/10.1080/00140130500070921>
- Costigan, P.A., Deluzio, K.J., Wyss, U.P., 2002. Knee and hip kinetics during normal stair climbing. *Gait Posture* 16, 33-37. [https://doi.org/10.1016/S0966-6362\(01\)00201-6](https://doi.org/10.1016/S0966-6362(01)00201-6)
- Crowe, A., Schiereck, P., Keessen, W., 1993. Gait adaptations of young adult females to hand-held loads determined from ground reaction forces. *Gait Posture* 1, 154-160. [https://doi.org/10.1016/0966-6362\(93\)90057-8](https://doi.org/10.1016/0966-6362(93)90057-8)
- de Kam, D., Kamphuis, J.F., Weerdesteyn, V., Geurts, A.C.H., 2016. The effect of weight-bearing asymmetry on dynamic postural stability in healthy young individuals. *Gait Posture* 45, 56-61. <https://doi.org/10.1016/j.gaitpost.2016.01.004>
- DeVita, P., Hong, D., Hamill, J., 1991. Effects of asymmetric load carrying on the biomechanics of walking. *J. Biomech.* 24, 1119-29. [https://doi.org/10.1016/0021-9290\(91\)90004-7](https://doi.org/10.1016/0021-9290(91)90004-7)
- DiDomenico, A., McGorry, R.W., Banks, J.J., 2015. Factors affecting time-to-contact

calculations during quiet standing. *Motor Control* 19, 1–9. <https://doi.org/10.1123/mc.2013-0019>

Fowler, N.E., Rodacki, A.L.F., Rodacki, C.D., 2006. Changes in stature and spine kinematics during a loaded walking task. *Gait Posture* 23, 133–41. <https://doi.org/10.1016/j.gaitpost.2004.12.006>

Fraga, M., Cristina, G., Noll, M., Barbosa, F., Silveira, I., Avelar, D., Hentschel, P., 2016. Use of a backpack alters gait initiation of high school students. *J. Electromyogr. Kinesiol.* 28, 82–89. <https://doi.org/10.1016/j.jelekin.2016.03.008>

Gök, H., Ergin, S., Yavuzer, G., 2002. Kinetic and kinematic characteristics of gait in patients with medial knee arthrosis. *Acta Orthop. Scand.* 73, 647–52. <https://doi.org/10.1080/000164702321039606>

Haahr, M., 1998. List Randomizer. <http://www.random.org/lists/>.

Haddad, J.M., Gagnon, J.L., Hasson, C.J., Van Emmerik, R.E.A., Hamill, J., 2006. Evaluation of time-to-contact measures for assessing postural stability. *J. Appl. Biomech.* 29, 111–8. <https://doi.org/10.1123/jab.22.2.155>

Hertel, J., Olmsted-Kramer, L.C., 2007. Deficits in time-to-boundary measures of postural control with chronic ankle instability. *Gait Posture* 25, 33–9. <https://doi.org/10.1016/j.gaitpost.2005.12.009>

Hill, M.W., Duncan, M.J., Oxford, S.W., Kay, A.D., Price, M.J., 2018. Effects of external loads on postural sway during quiet stance in adults aged 20–80 years. *Appl. Ergon.* 66, 64–69. <https://doi.org/10.1016/j.apergo.2017.08.007>

- Hong, Y., Li, J.X., 2005. Influence of load and carrying methods on gait phase and ground reactions in children's stair walking. *Gait Posture* 22, 63–8.  
<https://doi.org/10.1016/j.gaitpost.2004.07.001>
- Krebs, D.E., Wong, D., Jevsevar, D., Riley, P.O., Hodge, W.A., 1992. Trunk kinematics during locomotor activities. *Phys. Ther.* 72, 505–14. <https://doi.org/10.1093/ptj/72.7.505>
- Liberty Mutual Research Institute for Safety, 2019. Liberty Mutual Research Institute for Safety 2019 Workplace Safety Index. [/http://www.libertymutual.com/researchinstitute](http://www.libertymutual.com/researchinstitute)
- Lin, Y.C., Fok, L.A., Schache, A.G., Pandy, M.G., 2015. Muscle coordination of support, progression and balance during stair ambulation. *J. Biomech.* 48, 340–7.  
<https://doi.org/10.1016/j.jbiomech.2014.11.019>
- Liu, J., Lockhart, T.E., 2013. Local dynamic stability associated with load carrying. *Saf. Health Work* 4, 46–51. <https://doi.org/10.5491/SHAW.2013.4.1.46>
- Madehkhaksar, F., Egges, A., 2016. Effect of dual task type on gait and dynamic stability during stair negotiation at different inclinations. *Gait Posture* 43, 114–9.  
<https://doi.org/10.1016/j.gaitpost.2015.09.009>
- Maki, B.E., 1997. Gait changes in older adults: Predictors of falls or indicators of fear? *J. Am. Geriatr. Soc.* 45, 313–20. <https://doi.org/10.1111/j.1532-5415.1997.tb00946.x>
- Maki, B.E., Holliday, P.J., Topper, A.K., 1994. A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *Journals Gerontol.* 49, M72-M84. <https://doi.org/10.1093/geronj/49.2.M72>
- Matsuo, T., Hashimoto, M., Koyanagi, M., Hashizume, K., 2008. Asymmetric load-carrying in

- young and elderly women: Relationship with lower limb coordination. *Gait Posture* 28, 517–20. <https://doi.org/10.1016/j.gaitpost.2008.02.001>
- Nadeau, S., McFadyen, B.J., Malouin, F., 2003. Frontal and sagittal plane analyses of the stair climbing task in healthy adults aged over 40 years: What are the challenges compared to level walking? *Clin. Biomech.* 18, 950–9. [https://doi.org/10.1016/S0268-0033\(03\)00179-7](https://doi.org/10.1016/S0268-0033(03)00179-7)
- Oates, A.R., Patla, A.E., Frank, J.S., Greig, M.A., 2005. Control of dynamic stability during gait termination on a slippery surface. *J. Neurophysiol.* 93, 64–70. <https://doi.org/10.1152/jn.00423.2004>
- Park, K., Sy, J.F., Horn, G.P., Kesler, R.M., Petrucci, M.N., Rosengren, K.S., Hsiao-Wecksler, E.T., 2018. Assessing gait changes in firefighters after firefighting activities and while carrying asymmetric loads. *Appl. Ergon.* 70, 44–50. <https://doi.org/10.1016/j.apergo.2018.01.016>
- Qu, X., 2013. Effects of cognitive and physical loads on local dynamic stability during gait. *Appl. Ergon.* 44, 455–8. <https://doi.org/10.1016/j.apergo.2012.10.018>
- Rodrigues, P., TenBroek, T., Van Emmerik, R., Hamill, J., 2014. Evaluating runners with and without anterior knee pain using the time to contact the ankle joint complexes' range of motion boundary. *Gait Posture* 39, 48-53. <https://doi.org/10.1016/j.gaitpost.2013.05.024>
- Rosker, J., Markovic, G., Sarabon, N., 2011. Effects of vertical center of mass redistribution on body sway parameters during quiet standing. *Gait Posture* 33, 452–456. <https://doi.org/10.1016/j.gaitpost.2010.12.023>
- Slobounov, S., Cao, C., Sebastianelli, W., Slobounov, E., Newell, K., 2008. Residual deficits

- from concussion as revealed by virtual time-to-contact measures of postural stability. *Clin. Neurophysiol.* 119, 281-9. <https://doi.org/10.1016/j.clinph.2007.10.006>
- Slobounov, S.M., Moss, S.A., Slobounova, E.S., Newell, K.M., 1998. Aging and time to instability in posture. *Journals Gerontol.* 53, B71-8. <https://doi.org/10.1093/gerona/53A.1.B71>
- van Wegen, E.E.H., van Emmerik, R.E.A., Riccio, G.E., 2002. Postural orientation: age-related changes in variability and time-to-boundary. *Hum. Mov. Sci.* 21, 61–84. [https://doi.org/10.1016/S0167-9457\(02\)00077-5](https://doi.org/10.1016/S0167-9457(02)00077-5)
- Vera-Garcia, F.J., Brown, S.H.M., Gray, J.R., McGill, S.M., 2006. Effects of different levels of torso coactivation on trunk muscular and kinematic responses to posteriorly applied sudden loads. *Clin. Biomech. (Bristol, Avon)* 21, 443–55. <https://doi.org/10.1016/j.clinbiomech.2005.12.006>
- Wang, J., Gillette, J.C., 2018a. Carrying asymmetric loads during stair negotiation: Loaded limb stance vs. unloaded limb stance. *Gait Posture* 64, 213–9. <https://doi.org/10.1016/j.gaitpost.2018.06.113>
- Wang, J., Gillette, J.C., 2018b. Carrying asymmetric loads while walking on an uneven surface. *Gait Posture* 65, 39–44. <https://doi.org/10.1016/j.gaitpost.2018.06.173>
- Wang, K., Delbaere, K., Brodie, M.A.D., Lovell, N.H., Kark, L., Lord, S.R., Redmond, S.J., 2017. Differences between gait on stairs and flat surfaces in relation to fall risk and future falls. *IEEE J. Biomed. Heal. Informatics* 21, 1479-1486. <https://doi.org/10.1109/JBHI.2017.2677901>

Wheat, J.S., Haddad, J.M., Scaife, R., 2012. Between-day reliability of time-to-contact measures used to assess postural stability. *Gait Posture* 35, 345-7.

<https://doi.org/10.1016/j.gaitpost.2011.09.103>

Zultowski, I., Aruin, A., 2008. Carrying loads and postural sway in standing: the effect of load placement and magnitude. *Work* 30, 359–368. <https://doi.org/18725699>

Table 1. The effect of the loaded limb stance vs. the unloaded limb stance on COP parameters.

<i>Stair Ascent</i>			
	Loaded limb stance	Unloaded limb stance	<i>p</i>
Double stance ratio	0.472 (0.030)	0.478 (0.033)	0.116
ML COP excursion (cm)	2.02 (0.47)	1.57 (0.40)	0.001
ML COP velocity (cm/s)	5.56 (1.24)	5.10 (1.32)	0.238
Minimum ML TTC (s)	0.291 (0.050)	0.277 (0.058)	0.403
ML TTC percentage (%)	91.7 (2.9)	91.8 (4.5)	0.973
<i>Stair Descent</i>			
	Loaded limb stance	Unloaded limb stance	<i>p</i>
Double stance ratio	0.382 (0.039)	0.395 (0.041)	0.010
ML COP excursion (cm)	2.77 (1.00)	2.44 (1.04)	0.125
ML COP velocity (cm/s)	9.10 (3.39)	7.41 (2.78)	0.008
Minimum ML TTC (s)	0.192 (0.053)	0.228 (0.059)	0.011
ML TTC percentage (%)	78.5 (8.0)	85.0 (8.1)	0.001

Table 2. Mean (standard deviation) COP parameters during stair ascent and descent

<i>Stair ascent</i>				
	No load Mean (SD)	Bilateral Mean (SD)	Unilateral Mean (SD)	Main effect <i>p</i>
Double stance ratio	0.434 (0.030)	0.460 <sup>a</sup> (0.027)	0.470 <sup>a</sup> (0.034)	<0.001*
ML COP excursion (cm)	1.62 (0.33)	1.90 <sup>a</sup> (0.49)	2.04 <sup>a</sup> (0.46)	<0.001*
ML COP velocity (cm/s)	4.48 (0.71)	5.56 <sup>a</sup> (1.22)	6.09 <sup>a</sup> (1.24)	<0.001*
ML TTC percentage (%)	91.4 (3.0)	88.1 <sup>a</sup> (4.9)	87.5 <sup>a</sup> (3.5)	<0.001*
Minimum ML TTC (s)	0.284 (0.039)	0.253 <sup>a</sup> (0.037)	0.236 <sup>a</sup> (0.037)	<0.001*
<i>Stair descent</i>				
	No load Mean (SD)	Bilateral Mean (SD)	Unilateral Mean (SD)	Main effect <i>p</i>
Double stance ratio	0.335 (0.029)	0.361 <sup>a</sup> (0.042)	0.365 <sup>a</sup> (0.042)	0.002*
ML COP excursion (cm)	2.53 (0.72)	2.74 (0.85)	2.93 <sup>a</sup> (1.00)	0.011*
ML COP velocity (cm/s)	7.96 (2.28)	8.57 (2.62)	9.54 <sup>a</sup> (3.30)	<0.001*
ML TTC percentage (%)	77.5 (6.1)	78.0 (4.7)	74.8 <sup>b</sup> (7.6)	0.002*
Minimum ML TTC (s)	0.185 (0.033)	0.186 (0.028)	0.165 <sup>ab</sup> (0.040)	0.022*

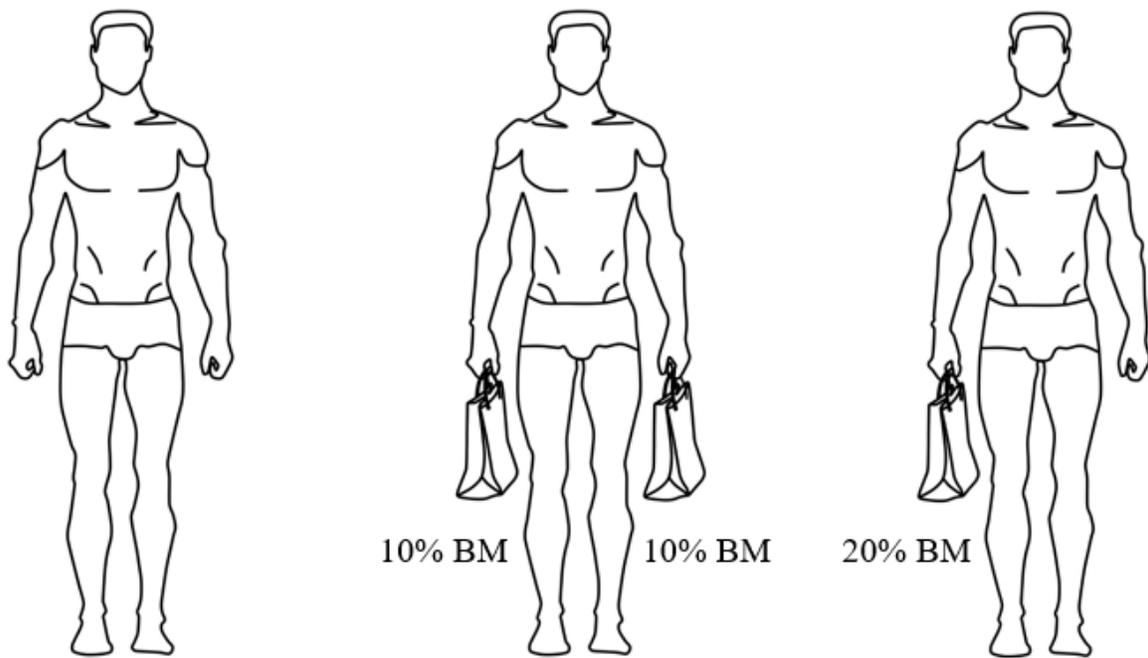
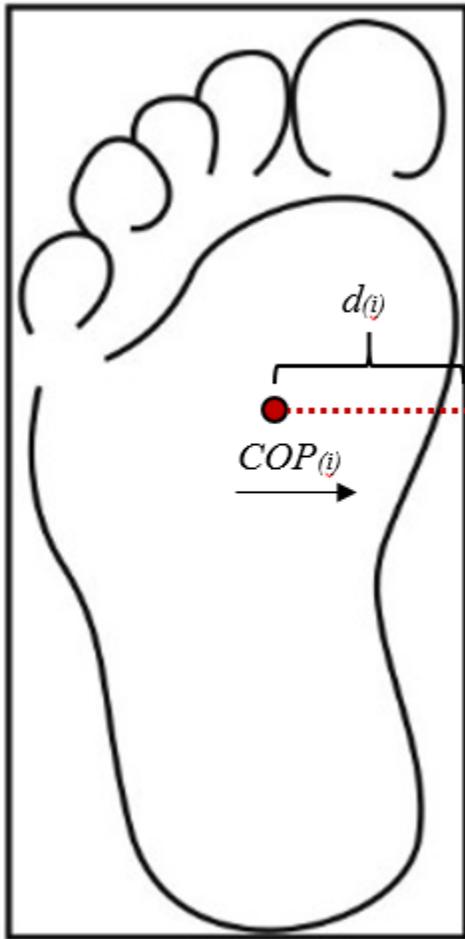


Figure 1. Illustration of the three load conditions: no load (left), 20% BM bilateral load split between both sides of the body (center), and 20% BM unilateral load on one side of the body (right).



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

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

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

$i$  : each data point ( $\frac{1}{160}$  s interval)

Figure 2. Illustration of the rectangular boundary of the foot and the Time-to-Contact calculation.  $v$  and  $a$  (COP velocity and acceleration) were calculated from the  $COP_{(i)}$  using the first central difference method.

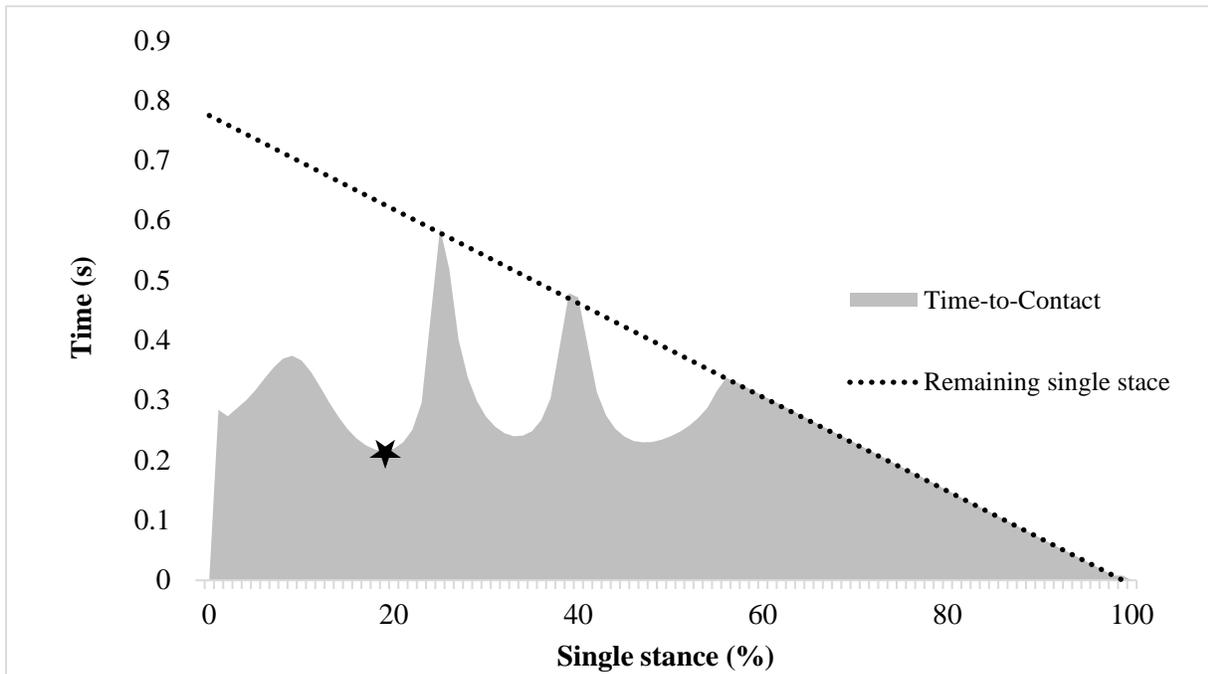


Figure 3. Illustration of dynamic TTC: if TTC is less than remaining single stance time, then TTC is saved. If TTC is greater than remaining single stance time, then TTC is equal to remaining single stance time (no adjustment needed). Star represents minimum dynamic TTC. TTC percentage was calculated by the ratio of mean TTC and mean remaining single stance time during the single stance phase (100% indicates no adjustments needed).