



# Untangling biomechanical differences in perturbation-induced stepping strategies for lateral balance stability in older individuals

J. Borrelli<sup>a,\*</sup>, R. Creath<sup>b</sup>, V.L. Gray<sup>a</sup>, M.W. Rogers<sup>a</sup>

<sup>a</sup> University of Maryland School of Medicine, MD, USA

<sup>b</sup> Lebanon Valley College, PA, USA

## ARTICLE INFO

### Article history:

Accepted 27 November 2020

### Keywords:

Aging  
Falls  
Lateral balance  
Stepping  
Perturbation  
Rehabilitation

## ABSTRACT

When recovering balance from a lateral perturbation, younger adults tend to stabilize balance with a single lateral sidestep while older adults often take multistep responses. Using multiple steps to recover balance is consistently associated with increased fall risk, altered body center of mass (CoM) control and instability. The aim of this study was to compare the spatio-temporal stepping characteristics and the margin of stability (MoS) of single lateral sidesteps (LSS1) with the first and second steps of a two-step protective step sequence. Two-step sequences begin with either a cross-over step to the front or back, or a medial step followed by a lateral sidestep. Seventy-one older adults received random lateral waist-pull perturbations to either side. We hypothesized that LSS1 would be more stable (larger MoS) than either step in a two-step sequence. With some exceptions, utilizing a two-step sequence was associated with a reduced CoM velocity and distance between the base of support and CoM and decreased stability in the frontal plane following limb loading of the first and second step. There were no differences in the time available to arrest the extrapolated CoM at the end of a single lateral sidestep or the final step of a two-step sequence. Two-step sequences involving a cross-over step include more complex stepping trajectories and also challenge stability in the sagittal plane requiring a multidimensional balance correction. These results indicate important step type differences in center of mass control in recovering balance with a single lateral sidestep as opposed to a two-step sequence among older adults.

© 2020 Elsevier Ltd. All rights reserved.

## 1. Introduction

Risk for hip fracture in older adults is six times greater during sideways falls than forward or backward falls (Nevitt and Cummings, 1993). Many older adults have impaired weight transfer control and difficulties stepping sideways as a protective response to a lateral loss of balance (Mille et al., 2005; Rogers and Mille, 2003). These difficulties have been proposed to contribute to decreased stability of the first step necessitating additional protective steps (Borrelli et al., 2019; Fujimoto et al., 2017; Mille et al., 2013). The need for multiple steps is associated with

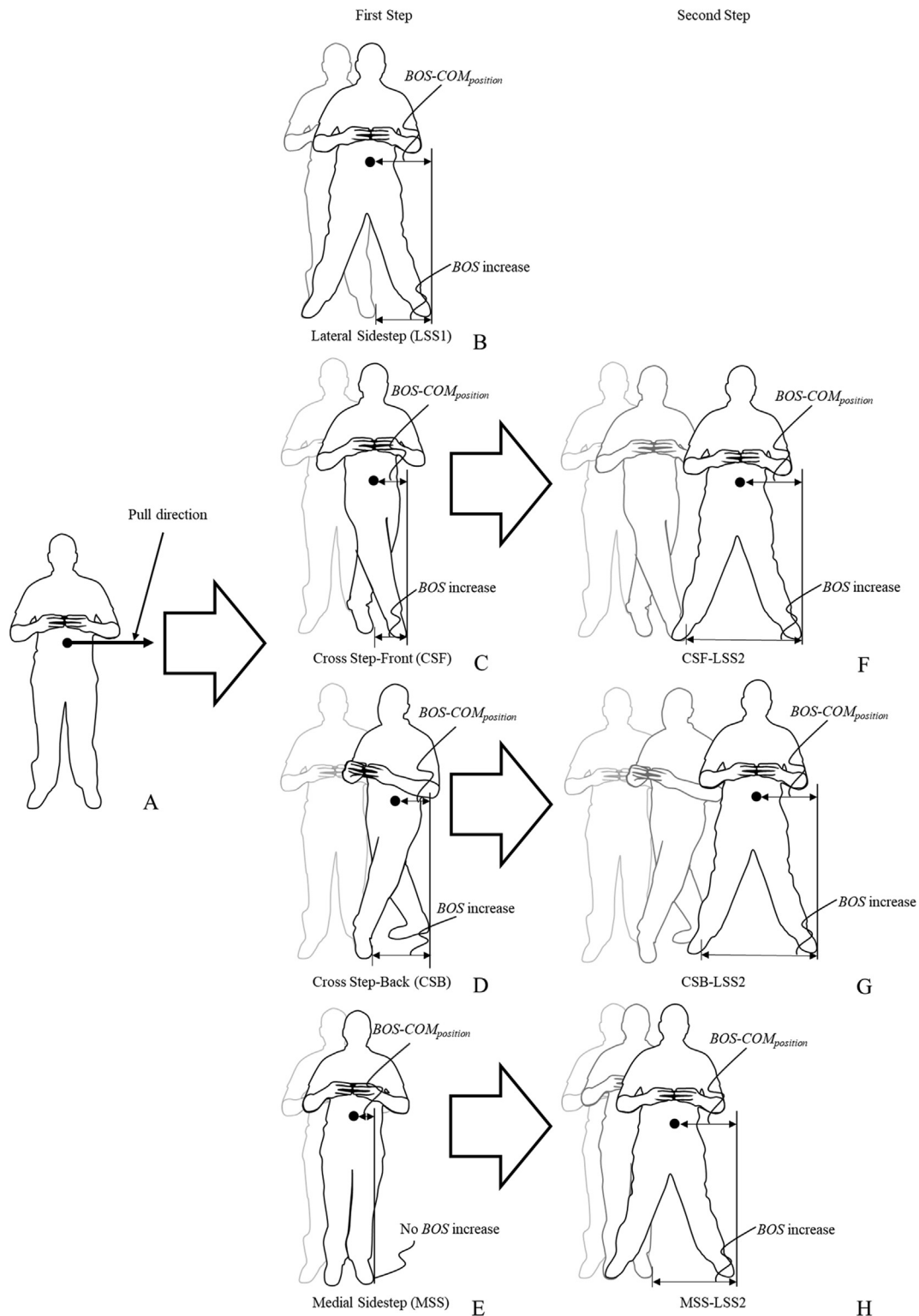
an increased risk of falling in older adults (Johnson-Hilliard et al., 2010). While many studies have investigated the first step of single and multiple protective stepping reactions (e.g. Bair et al., 2016; Hurt et al., 2011; Maki et al., 2000), little is known about the spatio-temporal stepping patterns and balance stability of the second step of a multistep stepping pattern.

Balance recovery through protective stepping requires regulation of the spatial and temporal relationship between the position and motion of the base of support (BoS) and the center of mass (CoM, Mille et al., 2005). Following lateral perturbations during stationary standing, the CoM is moved passively, relative to the BoS, such that the leg contralateral to the direction of the imposed CoM movement is passively unloaded and the ipsilateral leg is passively loaded (Maki et al., 1996; Mille et al., 2005; Yungheer et al., 2012). The two main strategies involve stepping with either the passively loaded limb or the passively unloaded limb (Maki et al., 2000; Mille et al., 2005; Rogers and Mille, 2003) that are characterized by different spatio-temporal patterns. Steps initiated with the passively loaded limb are termed lateral sidesteps (LSS, Fig. 1B). LSS require a rapid shift of body weight support from the passively

**Abbreviations:** AB-AD, Hip abductor-adductor; AP, Anteroposterior; BoS, Base of Support; CoM, Center of Mass; CSB, Cross-over step to the front; CSF, Cross-over step to the back; EMM, Estimated marginal mean; LSS1, Single lateral sidestep; LSS2, Lateral sidestep following an unloaded sidestep; MoS, Margin of stability; MSS, Medially directed sidestep; ML, Mediolateral; SEM, Standard error of the mean; USS, Unloaded sidestep; XCoM, Extrapolated Center of Mass.

\* Corresponding author at: Room 108, 100 Penn St. Baltimore, MD 21201, USA.

E-mail addresses: [jrborrelli@gmail.com](mailto:jrborrelli@gmail.com) (J. Borrelli), [mwrogers@som.umaryland.edu](mailto:mwrogers@som.umaryland.edu) (M.W. Rogers).



**Fig. 1.** Following a lateral perturbation (A), participants stepped with the passively loaded leg resulting in a single lateral sidestep (LSS1, B) or they stepped with the passively unloaded leg resulting in an unloaded sidestep (C, D, E). There are three types of unloaded sidesteps (USS); a cross-over step to the front (CSF), a cross-over step to the back (CSB), and medial sidestep (MSS). Following an USS, a second step is usually required to recover balance. The step taken after an USS is a lateral sidestep which we have termed LSS2. Adapted from Borrelli et al., (2019) with permission from Elsevier.

loaded to the unloaded limb prior to step lift-off (Inacio et al., 2018). Steps initiated with the passively unloaded limb are termed unloaded sidesteps (USS, Fig. 1C, D, and E). There are three types of USS: cross-over step to the front (CSF, Fig. 1C), cross-over step to the back (CSB, Fig. 1D), and medially directed sidestep (MSS, Fig. 1E).

Younger adults primarily respond to a lateral loss of balance with a single lateral sidestep (LSS1), while older adults take more multiple steps starting with an USS (Mille et al., 2005). Differences in margin of stability (MoS, Hof et al., 2005) after the first protective step differentiates single and multiple protective stepping reactions (Borrelli et al., 2019; Fujimoto et al., 2017). The MoS is

smaller in single CSB and single MSS, but not single CSF, compared to LSS1 (Borrelli et al., 2019). Although single USS are also associated with a slower CoM velocity than LSS1, the increase in the BoS boundary is smaller and the CoM is closer to the BoS boundary compared to LSS1 (Borrelli et al., 2019). Additionally, multiple stepping reactions increase the risk of interlimb collision (Maki et al., 2000; Mille et al., 2005) and falls (Johnson-Hilliard et al., 2010).

When a second step is used following an USS, we propose calling the step sequence an USS followed by a lateral sidestep (USS-LSS2). The initial USS repositions the stance foot for the subsequent LSS2 compared to a LSS1, and may affect the spatio-temporal stepping characteristics and MoS of the LSS2 compared to a LSS1. However, it is not known if there are differences in the stepping characteristics, CoM control, and MoS between LSS1 and LSS2 which may affect fall risk.

While lateral stability is largely managed in the frontal plane, information on associated balance stability and stepping characteristics in the sagittal plane that accompany mediolateral (ML) balance recovery has been virtually unreported. Therefore, gaining further insight into the spatio-temporal stepping characteristics and stability of single and multistep reactions in the frontal and sagittal planes is warranted.

The purpose of this study was to compare the spatio-temporal stepping parameters and MoS between a single LSS1 and each step of a two-step (USS-LSS2) protective stepping sequence. We hypothesized that recovering balance using a LSS1 is associated with a greater MoS than multistep components in the frontal and sagittal planes. Furthermore, we expected LSS1 to have a larger BoS increase, larger distance between the BoS and CoM, and greater CoM velocity compared to USS and LSS2. Lastly, we hypothesized that LSS1 would be associated with smaller sagittal plane excursion of stepping parameters (CoM displacement, CoM velocity, step length, BoS increase).

## 2. Methods

### 2.1. Subjects

Seventy-one community-dwelling older adults were recruited from the greater Baltimore area, and from the Geriatric Research, Education and Clinical Center of the Baltimore Veterans Affairs Medical Center. Participant demographics are summarized in Table 1. Exclusion criteria included: 1) impaired cognitive function (Folstein et al., 1975); 2) depression (Sawyer Radloff, 1977); 3) sedative prescription; 4) significant impairment that limited functional activities; 5) non-ambulatory or used a walking aide; 6) were currently participating in vigorous exercise or muscle strengthening program; and 7) advised not to exercise by a physician. All participants provided written informed consent prior to participation and the study was approved by the Institutional

Review Board at the University of Maryland School of Medicine and the Baltimore Veterans Affairs Medical Center.

### 2.2. Testing protocol

Participants received 36 randomly applied, motor-driven lateral waist-pull perturbations (Pidcoe and Rogers, 1998). Three trials were conducted at each of six pull intensities (total displacement: 5, 8.6, 12.1, 15.7, 19.3, 22.8 cm; constant velocity: 9, 18, 27, 36, 45, 54 cm/s, and maximum allowable acceleration: 180, 360, 540, 720, 900, 900 cm/s<sup>2</sup>) in the left and right directions. Fig. 2 shows an example perturbation profile. The order of trial presentation was randomized to minimize anticipatory and sequence learning effects. A safety harness was worn to prevent contact with the ground due to loss of balance. Participants stood on two separate force platforms (AMTI, Watertown, MA) in a self-selected, comfortable standing position and held a light cylindrical baton with both hands in front of the body. The baton was used to minimize obstructing motion capture markers and prevent participants from grabbing the steel cables that were attached to the waist-belt they wore. Prior to testing, foot position was traced on contact paper to ensure consistent foot placement. Participants were instructed to “relax and react naturally to maintain your balance and prevent yourself from falling.”

### 2.3. Data collection and analysis

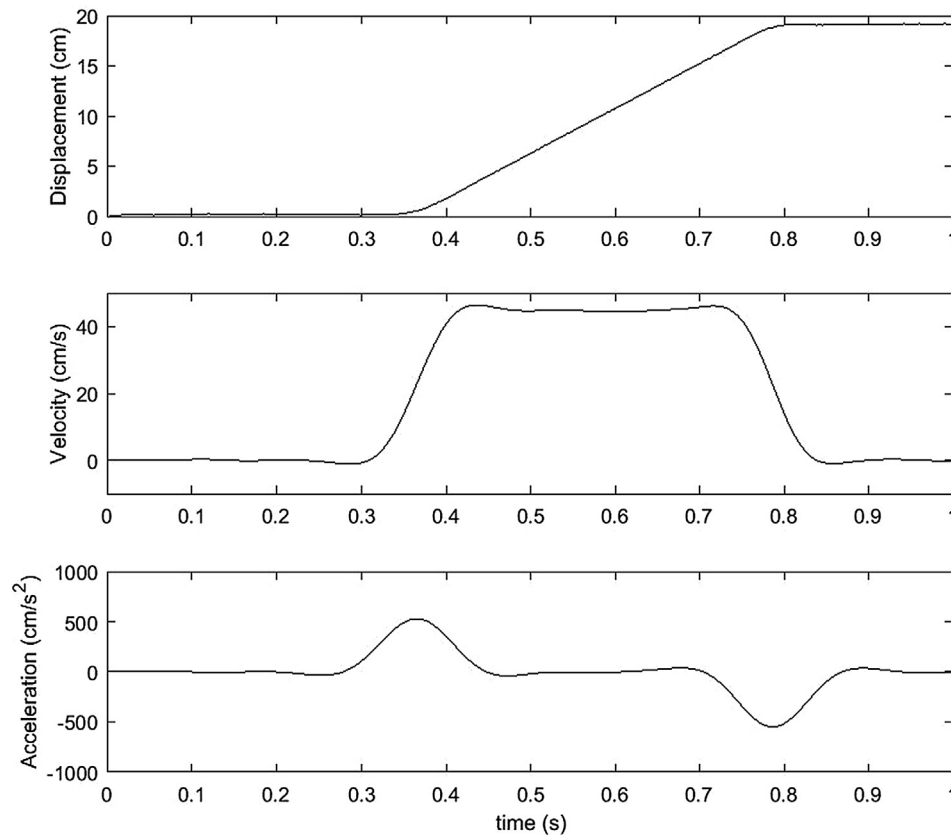
This study focused on single step and two step responses evoked by unpredictable lateral waist-pull perturbations. The perturbation characteristics that evoke stepping are relatively varied between individuals (Mille et al., 2003; Pai et al., 1998; Sturnieks et al., 2012). Analysis focused on the balance tolerance limit (Yungher et al., 2012), or the minimum pull intensity where multiple steps occurred (mean number of steps > 1) which was specific to each participant. There is a greater incidence of multiple steps in older than younger adults (Mille et al., 2013, 2005) which is well-associated with balance deficit and past and future-falls (Inacio et al., 2019; Johnson-Hilliard et al., 2010; Mille et al., 2013; Sturnieks et al., 2013). The minimum pull intensity where multiple steps occur is a threshold level related to fall risk (Sturnieks et al., 2013) and reflects “balance tolerance” across a range of balance perturbation intensities. The average (standard deviation) perturbation magnitude at the balance tolerance limit was  $4.7 \pm 1.2$  with the balance tolerance limit occurring at level 2, 3, 4, 5, and 6 for 11, 21, 20, 14, and 5 participants respectively. Analysis focused on 424 trials at the balance tolerance limit (2 participants only performed 5 trials at their balance tolerance limit).

Three-dimensional motion-capture (Vicon, Oxford, UK) was used to collect kinematic data. Motion capture data was sampled at 120 Hz and low-pass filtered at 6 Hz using a dual-pass fourth-order digital filter. Motion capture markers were placed at the bilateral posterior superior iliac spine, the lateral malleoli, the medial malleoli, 5th metatarsophalangeal joints, heel, and anterior to the 1st metatarsal head (toe).

Difficulties associated with lateral stepping prevented determination of stepping events (lift-off and touch down) using the force platforms (described further in Appendix A). As a result, stepping events were determined using vertical ankle velocity and we have termed the events step unloading onset and step loading as opposed to lift-off and touch down. Step unloading onset was defined as the time that the lateral malleolus marker vertical velocity exceeded 0.01 m/s. Step loading was defined as the time that the lateral ankle marker velocity was less than 0.01 m/s in the downward direction. The time between limb unloading onset and loading (limb unloading/loading phase), includes the stepping phase and an unloading phase prior to stepping and a loading

**Table 1**  
Participant demographics.

n	71
Females	38
Age (years)	72.7 (5.5)
Height (cm)	166.3 (8.7)
Weight (kg)	77.3 (16.2)
7-item Berg Balance Score <sup>1</sup>	26.8 (1.7)
Timed Up and Go <sup>2</sup> (s)	10.7 (1.9)
Physical Activity Scale for the Elderly <sup>3</sup>	116.4 (48.1)
Falls Efficacy Scale-International <sup>4</sup>	11.0 (1.9)
Activities-Specific Balance Confidence Scale <sup>5</sup>	93.5 (5.6)
4-item Dynamic Gait Index <sup>6</sup>	11.0 (1.3)
Four Square Step Test <sup>7</sup> (s)	9.4 (2.4)



**Fig. 2.** Representative data showing pull characteristics of displacement, velocity and acceleration of a trial with constant velocity of 45 cm, total displacement of 19.3 cm, and maximum acceleration of 900 cm/s<sup>2</sup>. Trials were conducted at six pull intensities (total displacement: 5, 8.6, 12.1, 15.7, 19.3, 22.8 cm; constant velocity: 9, 18, 27, 36, 45, 54 cm/s, and maximum allowable acceleration: 180, 360, 540, 720, 900, 900 cm/s<sup>2</sup>). The maximum allowable acceleration was not reached in any trial and the acceleration did not exceed 75% of the allowable maximum in any trial. The mean rise time for the velocity is 60 ms (Pidcoe and Rogers, 1998). Note that the deceleration of the cable pull is not transmitted to the participants due to the compliant nature of the cable used to transmit the perturbation to participants.

phase following stepping. It is shown in [Appendix A](#) that the unloading phase and loading phase, before lift-off and after touch down, take about 150 ms each. The time between limb unloading onset and loading (limb unloading/loading phase), BoS increase, distance between the CoM and BoS boundary (BoS-CoM) at step loading, CoM displacement at step loading, and CoM velocity at step loading were also calculated to assess stepping characteristics (see [Table 2](#) for a detailed description and [Fig. 3](#) for an example of data from a single trial).

Stability was quantified using the MoS. The MoS is defined as the difference between the BoS boundary and the extrapolated center of mass ([Hof et al., 2005](#)),

$$\text{MoS} = \text{BoS} - \text{XCoM}.$$

The BoS boundary was determined using markers on the feet (toe, lateral malleolus, heel, and 5th metatarsophalangeal joint marker). The extrapolated center of mass (XCoM) is the sum of the CoM position and the product of the CoM velocity and a scaling factor,

$$\text{XCoM} = \left( x + \dot{x} / \sqrt{g/L} \right)$$

where  $g$  is gravitational acceleration,  $L$  is the average body CoM height prior to perturbation, and  $x$  and  $\dot{x}$  are the CoM displacement and velocity. The CoM position was approximated at a point 0.17 m anterior to the midpoint of the posterior iliac spine markers ([Yang and Pai, 2014](#)).

The amount of time available,  $\tau$ , before the XCoM reaches the BOS, or the ‘time-to-contact’, was also calculated ([Hof et al.,](#)

[2005](#)). Dependent variables were evaluated in the ML and antero-posterior (AP) direction. Additionally, calculation of the change in angle of the feet and pelvis in the transverse plane are also described and reported in [Appendix B](#).

#### 2.4. Statistical analyses

Results are reported as estimated marginal means (EMM) and standard error of the mean (SEM). Statistical analysis was performed using SPSS v22 (IBM Corp, Armonk, NY). A linear mixed effects model with participants as a random factor, test number as a repeated within-participant factor, and step type as a fixed effect were used to compare step count between first step type (LSS1 and CSF, LSS1 and CSB, LSS1 and MSS).

Separate linear mixed effects models were used to compare the remaining dependent variables with participant as a random factor, test number as a repeated within-participant factor, and step type as a fixed effect. Only trials where balance was recovered with a single LSS1 or an USS followed by a LSS2 were included in the analyses. Linear mixed effects models were used to compare the dependent variables at first step loading and at step loading of the final balance recovery step of a LSS1 and CSF-LSS2, CSB-LSS2, and MSS-LSS2.

When first step type (between LSS1, CSF, CSB, and MSS) had a significant main effect in the analysis, post-hoc comparisons between step types were evaluated using Fisher’s least significant differences. When step type had a significant main effect on final step loading (LSS1 and LSS2), post-hoc comparisons using Fisher’s least significant differences were performed between single LSS1

**Table 2**  
Description of dependent variables, their abbreviations, and units.

Variable	Abbreviation	Units	Description
Step Unloading Onset		milliseconds	Lateral ankle marker velocity was greater than 0.01 m/s in upward direction.
Limb Unloading/Loading Phase		milliseconds	Difference between the step loading and unloading onset of the stepping leg. This time period includes unloading of the swing leg, the swing phase, contact, and loading of the swing leg.
Step Loading		milliseconds	Lateral ankle marker velocity was less than 0.01 m/s in the downward direction.
Mediolateral Step Length	ML Step Length	Percent Height (% height)	Distance between the lateral malleolus at first or second step unloading onset and first or second step loading respectively, in the frontal plane.
Anteroposterior Step Length	AP Step Length	Percent Height (% height)	Distance between the lateral malleolus at first or second step unloading onset and first or second step loading in the sagittal plane respectively.
Mediolateral Base of Support Increase	ML BoS Increase	Percent Height (% height)	The change in base of support boundary (in the presumed fall direction) at first or second step loading compared to first or second step unloading onset respectively in the frontal plane
Mediolateral Base of Support			The most lateral aspects of the foot (toe, fifth metatarsophalangeal joint, heel, lateral malleolus, medial malleolus) in the presumed fall direction in the frontal plane.
Anteroposterior Base of Support Increase	AP BoS Increase	Percent Height (% height)	The change in base of support boundary between step loading (first or second) and first step unloading onset in the sagittal plane
Anteroposterior Base of Support			Distance between most anterior and posterior aspects of the feet (toe, fifth metatarsophalangeal joint, heel, lateral malleolus, and medial malleolus in the sagittal plane.
Mediolateral Center of Mass Displacement	ML CoM Displacement	Percent Height (% height)	Difference between the center of mass position at first or second step loading and first or second step unloading onset respectively in the frontal plane.
Anteroposterior Center of Mass Displacement	AP CoM Displacement	Percent Height (% height)	Difference between the center of mass position at first or second step loading and first or second step unloading onset respectively in the sagittal plane.
Mediolateral Distance between the Base of Support and Center of Mass	ML BoS-CoM Distance	Percent Height (% height)	Distance between the center of mass and base of support boundary in the presumed fall direction in the frontal plane at first or second step loading.
Anterior Distance between the Base of Support and Center of Mass	Anterior BoS-CoM Distance	Percent Height (% height)	Distance between the center of mass and anterior base of support in the sagittal plane at first or second step loading.
Posterior Distance between the Base of Support and Center of Mass	Posterior BoS-CoM Distance	Percent Height (% height)	Distance between the center of mass and posterior base of support in the sagittal plane at first or second step loading.
Mediolateral Center of Mass Velocity	ML CoM Velocity	Percent Height/s (% height/s)	Center of mass velocity in the frontal plane at first or second step loading.
Anteroposterior Center of Mass Velocity	AP CoM Velocity	Percent Height/s (% height/s)	Center of mass velocity in the sagittal plane at first or second step loading.
Extrapolated Center of Mass	XCoM	Percent Height (% height)	Sum of the center of mass position and the product of the center of mass velocity and the scaling factor.
Scaling factor		Seconds (s)	Square root of the average center of mass height prior to step unloading onset divided by the acceleration due to gravity.
Mediolateral Margin of Stability	ML MoS	Percent Height (% height)	Difference between the base of support boundary and the extrapolated center of mass at first or second step loading.
Anterior Margin of Stability	Anterior MoS	Percent Height (% height)	Difference between the base of support boundary and the extrapolated center of mass with respect to the anterior base of support boundary limit at first or second step loading.
Posterior Margin of Stability	Posterior MoS	Percent Height (% height)	Difference between the base of support boundary and the extrapolated center of mass with respect to the posterior base of support boundary limit at first or second step loading.
Time it will take the extrapolated center of mass to reach the base of support boundary	$\tau$	Milliseconds (ms)	The relationship between center of mass velocity and margin of stability reflects the amount of time available to decrease the center of mass velocity in order to prevent the center of mass from reaching the base of support boundary limit (Hof et al., 2005). The amount of time available is equal to the ratio of margin of stability over center of mass velocity (Hof et al., 2005).

and LSS2 following a CSF (CSF-LSS2), LSS1 and LSS2 following a CSB (CSB-LSS2), and LSS1 and LSS2 following a MSS (MSS-LSS2).

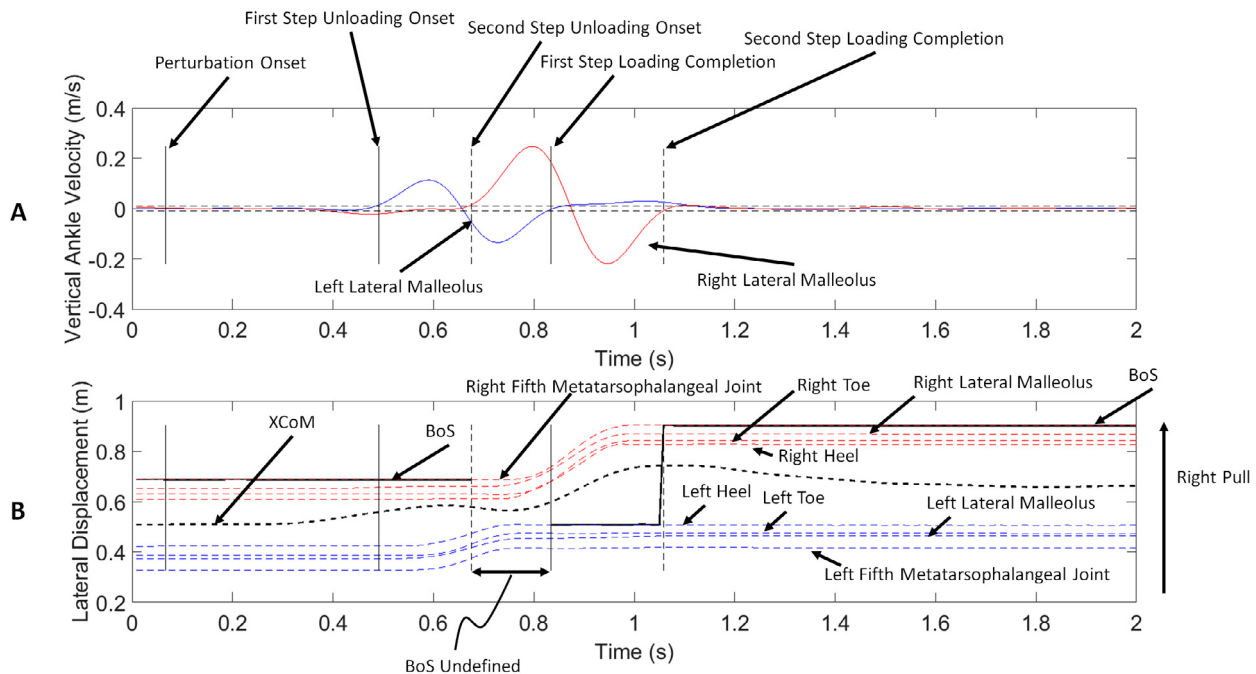
### 3. Results

Table 3 lists the average number of steps taken to recover balance when the first step was a LSS, CSF, CSB, or MSS, and the num-

ber of trials with no steps, one step, two steps or more than two steps. Step type had a significant effect on the number of recovery steps taken ( $p < 0.001$ ). Post-hoc analysis ( $\alpha = 0.05$ ) showed that fewer steps were required to recover balance when a LSS was taken as the first step.

Balance was recovered with a LSS1 (18%, 70/398) or a LSS2 (46%, 183/398) following an USS. Trials where the limb was raised and





**Fig. 3.** Representative data from a single trial of a right waist-pull perturbation resulting in a medial sidestep (MSS) followed by a lateral sidestep (LSS2). Panel A shows the vertical velocity of the left (blue) and right (red) lateral malleolus marker. The first step was performed with the left leg (blue line) and the second step was performed with the right leg (red line). Step unloading onset was defined as the time when the lateral malleolus vertical velocity was greater than 0.1 m/s (shown with a horizontal dashed line). Step loading was defined as the time when the vertical velocity of the lateral malleolus exceeded  $-0.1$  m/s (horizontal dashed line). Panel B shows the lateral position of the motion capture markers on the toe, heel, lateral malleolus, and 5th metatarsophalangeal joints of the left (blue) and right (red). The lateral boundary of the base of support (BoS) was defined as the most lateral marker position of the foot/feet that is/are substantially loaded. The extrapolated center of mass (XCoM) is the sum of the center of mass (CoM) position and the product of the CoM velocity and a scaling factor. Note that there is a brief period where body weight is being shifted to the landing limb. During this time, both feet were undergoing loading or unloading (first step loading after second step unloading onset) and the BoS was undefined. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

**Table 3**

Estimated marginal mean step count (standard error of the mean) for trials where a lateral sidestep (103), a cross-over step to the front (47), a cross-over step to the back (100), or a medially directed sidestep (148) were the first step type. The table also details the number of trials where no step, 1 step, 2 steps, or more than 2 steps were used to recover balance using a lateral sidestep, cross-over step to the front, cross-over step to the back, or medial sidestep. Trials where participants raised and lowered a leg without advancing the limb laterally were termed “flamingo” steps. Flamingo and no step trials were not included in the analysis. The highlighted trials consist of a single lateral sidestep (LSS1) and unloaded sidestep (cross-over step to the front, cross-over step to the back, or medially directed sidestep) followed by a lateral sidestep. The highlighted cells are the focus of the analysis. <sup>†</sup>Significant difference between lateral sidesteps and cross-over steps to the front ( $p = 0.013$ ). <sup>‡</sup>Significant difference between lateral sidestep and cross-over steps to the back ( $p < 0.001$ ). <sup>§</sup>Significant difference between lateral sidesteps and medial sidesteps ( $p < 0.001$ ).

First Step type	Step count	No steps	1 step	2 steps	> 2 steps	Total
Lateral sidestep	1.33 (0.06)	0	70	27	6	103
Cross-over step to the front	1.58 (0.09) <sup>†</sup>	0	19	27	1	47
Cross-over step to the back	1.65 (0.06) <sup>‡</sup>	0	40	45	15	100
Medially directed sidestep	1.94 (0.06) <sup>§</sup>	0	18	111	19	148
Flamingo	1.05 (0.20)	0	17	1	0	18
No step	-	8	0	0	0	8
Total		8	164	211	41	424

lowered but did not advance laterally (limb elevation or “flamingo” steps) or no step was taken were not analyzed. The remaining results focus on the step characteristics of trials where a LSS1 or USS-LSS2 were evoked. The trials of interest (64%; 253/398 trials) are highlighted in Table 3. Thirty-six trials were excluded from analysis because of excessive marker obstructions, leaving 217/253 trials (62 LSS1 trials, 20 CSF-LSS2 trials, 41 CSB-LSS2 trials, 94 MSS-LSS2 trials). Although 16/71 participants used only one type of stepping strategy (LSS, CSF, CSB, or MSS) in all trials, the majority (55/71) of participants used 2 or more stepping strategies

(1/71 used 4 strategies; 15/71 used 3 strategies; 39/71 used 2 strategies).

### 3.1. Mediolateral LSS1 and USS characteristics at first step loading

All dependent variables at first step loading were significantly affected by the main effect of step type ( $p$ -values  $< 0.001$ , Table 4). Step unloading onset occurred earlier for some USS (CSF: 340 ms,  $p = 0.009$ ; MSS: 345 ms,  $p < 0.001$ ) compared to LSS1 (408 ms). Step length for cross-over steps was longer for cross-over steps (CSF:

**Table 4**

Estimated marginal mean (EMM) and standard error of the mean (SEM) for step unloading onset and various outcome measures at step loading of the first step. *P*-values from post-hoc comparisons between a single lateral sidestep (LSS1) and cross-over step to the front (CSF), LSS1 and a cross-over step to the back (CSB), and LSS1 and a medial sidestep (MSS) are also shown. The number of trials (*n*) used in the calculation of the time it will take the extrapolated center of mass available to reach the base of support,  $\tau$ , is noted under each step type. In some trials the CoM velocity was in the direction opposite of the BoS boundary and the  $\tau$  is not defined. For example, in 1/62 LSS1 the CoM was moving away from the BoS boundary and  $\tau$  is not defined. \*Significant step type effect ( $p < 0.001$ ). #There was little variation from trial to trial and the model would not converge. A model with participant as a random effect and step type as a fixed effect was used.

Mediolateral First Step Characteristics	LSS1	CSF	p-value	CSB	p-value	MSS	p-value
	EMM (SEM)	EMM (SEM)		EMM (SEM)		EMM (SEM)	
Step Unloading Onset (ms)*	408 (14)	340 (23)	<b>0.009</b>	369 (16)	0.05	345 (11)	<b>&lt;0.001</b>
Limb Unloading/Loading Phase (ms)*	469 (25)	853 (43)	<b>&lt;0.001</b>	783 (26)	<b>&lt;0.001</b>	423 (21)	0.06
ML Step Length (% height)*	23.8 (1.1)	30.5 (1.6)	<b>&lt;0.001</b>	30.1 (1.2)	<b>&lt;0.001</b>	7.7 (0.9)	<b>&lt;0.001</b>
ML BoS Increase (% height)*	22.8 (1.0)	17.1 (1.5)	<b>0.001</b>	13.7 (1.2)	<b>&lt;0.001</b>	-10.1 (0.8)	<b>&lt;0.001</b>
ML CoM Displacement (% height)*	12.1 (0.7)	14.8 (1.1)	<b>0.031</b>	18.3 (0.8)	<b>&lt;0.001</b>	5.4 (0.6)	<b>&lt;0.001</b>
ML BoS-CoM Distance (% height)*	19.0 (0.7)	13.6 (1.1)	<b>&lt;0.001</b>	6.0 (0.8)	<b>&lt;0.001</b>	-6.0 (0.6)	<b>&lt;0.001</b>
ML CoM Velocity (% height/s)*	23.4 (1.7)	15.7 (2.6)	<b>0.006</b>	20.6 (1.9)	0.17	14.4 (1.5)	<b>&lt;0.001</b>
ML MoS (% height)*	12.3 (0.9)	8.9 (1.3)	<b>0.022</b>	-0.4 (1.1)	<b>&lt;0.001</b>	-10.2 (0.8)	<b>&lt;0.001</b>
$\tau$ (ms)#,*	1013 (153)	741 (207)	0.17	254 (170)	<b>&lt;0.001</b>	182 (145)	<b>&lt;0.001</b>
<i>n</i>	61/62	20/20		41/41		92/94	

31% height; CSB: 30% height) and shorter for MSSs (8% height) than LSS1 (24% height;  $p$ -values  $< 0.001$ ). Despite increased step length for CSF and CSB, the BoS increase was smaller for all USS compared to LSS1 ( $p$ -values  $< 0.001$ ). The CoM position was farther from the BoS boundary (BoS-CoM distance) at step loading in LSS1 (19% height) compared to USS ( $<14\%$  height,  $p$ -values  $< 0.001$ ). Although, the CoM velocity at step loading was smaller for CSF (16% height/s,  $p = 0.006$ ) and MSS (14% height/s,  $p < 0.001$ ) compared to LSS1 (23% height/s), the MoS at step loading was smaller for USS (CSF: 9% height,  $p = 0.022$ ; CSB: 0% height,  $p < 0.001$ ; MSS: -10% height,  $p < 0.001$ ) compared to LSS1 (12% height). Lastly, CSB (254 ms,  $p < 0.001$ ) and MSS (182 ms;  $p < 0.001$ ) had a significantly shorter  $\tau$  than LSS1 (1013 ms).

### 3.2. Mediolateral LSS1 and LSS2 characteristics at final step loading

First step type significantly affected the biomechanical characteristics of LSS2 compared to LSS1 ( $p$ -values  $< 0.001$ , Table 5). Limb unloading/loading took more than 40 ms longer in LSS2 compared to LSS1 ( $p$ -values  $< 0.004$ ) but resulted in smaller BoS increase (LSS1: 21% height; USS-LSS2:  $<17\%$  height,  $p$ -values  $< 0.001$ ). The distance between the BoS and CoM (BoS-CoM distance) was also larger for LSS1 (18% height) compared to LSS2 of a CSB-LSS2 (8% height,  $p < 0.001$ ) and LSS2 of MSS-LSS2 (15% height,  $p < 0.001$ ). Although the CoM velocity was larger for LSS1 (24% height/s) than LSS2 following an USS ( $<19\%$  height/s,  $p$ -values  $< 0.004$ ), the MoS was larger for LSS1 (11% height) compared to most LSS2 (CSB-

LSS2: 6% height,  $p < 0.001$ ; MSS: 9% height,  $p = 0.02$ ). Although the time it took the XCoM to reach the BoS boundary,  $\tau$ , was significantly affected by a main effect of step type, post-hoc analysis failed to identify any significant differences ( $p$ -values  $> 0.32$ ).

### 3.3. Anteroposterior stepping characteristics

AP step length, absolute AP BoS increase, and absolute AP CoM velocity were larger for CSF-LSS2 and CSB-LSS2 compared to LSS1. In general anterior and posterior MoS was larger for LSS1 compared to USS at first step loading and smaller when compared to LSS2 following a USS at step loading. There were no significant differences in  $\tau$ . For further details, see Appendix A2 and Tables A5 and A6.

## 4. Discussion

We hypothesized that a single lateral sidestep (LSS1) would have a larger MoS, BoS increase, BoS-CoM distance, and greater CoM velocity at step loading compared to the first and final step of an USS-LSS2. Additionally, we hypothesized LSS1 to have smaller excursion and increased MoS in the sagittal plane. The results largely supported our hypotheses.

Balance recovery requires regulating the CoM - BoS relationship (Hof, 2007; Mille et al., 2005; Pai and Patton, 1997). While both LSS1 and USS-LSS2 steps resolved the perturbation-induced instability, their stepping patterns differed significantly. LSS1 were

**Table 5**

Estimated marginal mean (EMM) and standard error of the mean (SEM) for step unloading onset and various dependent variables at final step loading. Dependent variables were calculated at the loading of the first step for single lateral sidesteps (LSS1) and at loading of the lateral sidestep (LSS2) following a cross-over step to the back (CSF-LSS2), cross-over step to the front (CSF-LSS2), and medial sidestep (MSS-LSS2). *P*-values from post-hoc comparisons between LSS1 and CSF-LSS2, LSS and CSB-LSS2, LSS and MSS-LSS2 are also shown. The number of trials (*n*) used in calculation of the time it will take the extrapolated center of mass (XCoM) available to reach the base of support,  $\tau$ , is noted under each step type. In some trials the CoM velocity was in the direction opposite of the BoS boundary and the  $\tau$  is not defined. For example, in 3/20 CSF-LSS2 the CoM was moving away from the BoS boundary and  $\tau$  is not defined. \*Significant step type effect ( $p < 0.001$ ).

Mediolateral LSS1 and LSS2 Characteristics	LSS1	CSF-LSS2	p-value	CSB-LSS2	p-value	MSS-LSS2	p-value
	EMM (SEM)	EMM (SEM)		EMM (SEM)		EMM (SEM)	
Step Unloading Onset (ms)*	456 (50)	1337 (82)	<b>&lt;0.001</b>	1094 (59)	<b>&lt;0.001</b>	658 (42)	<b>&lt;0.001</b>
Limb Unloading/Loading Phase (ms)*	440 (18)	569 (28)	<b>&lt;0.001</b>	641 (21)	<b>&lt;0.001</b>	499 (16)	<b>0.003</b>
ML Step Length (% height)*	21.1 (1.2)	20.0 (1.9)	0.59	19.7 (1.4)	0.38	18.1 (1.1)	<b>0.016</b>
ML BoS Increase (% height)*	21.1 (1.0)	9.5 (1.7)	<b>&lt;0.001</b>	8.1 (1.2)	<b>&lt;0.001</b>	17.2 (0.9)	<b>&lt;0.001</b>
ML CoM Displacement (% height)*	10.8 (0.8)	7.2 (1.2)	<b>0.006</b>	12.7 (1.0)	0.09	9.2 (0.8)	<b>0.041</b>
ML BoS-CoM Distance (% height)*	18.2 (0.6)	16.6 (1.0)	0.16	7.8 (0.7)	<b>&lt;0.001</b>	15.2 (0.5)	<b>&lt;0.001</b>
ML CoM Velocity (% height/s)*	23.7 (1.5)	16.5 (2.3)	<b>0.003</b>	7.7 (1.7)	<b>&lt;0.001</b>	19.2 (1.3)	<b>0.002</b>
ML MoS (% height)*	10.9 (0.6)	11.9 (1.1)	0.41	5.5 (0.8)	<b>&lt;0.001</b>	9.3 (0.6)	<b>0.02</b>
$\tau$ (ms)*	720 (112)	860 (173)	0.45	713 (123)	0.96	621 (84)	0.32
<i>n</i>	61/62	17/20		34/41		94/94	

more effective in increasing the BoS boundary through stepping. LSS1 directly increased the BoS boundary, whereas USS and LSS2 following CSF and CSB must clear the stance limb before the BoS boundary is increased. CSF-LSS2 and CSB-LSS2 are further complicated by more complex ML-AP step trajectories than LSS1. Additionally, the ML BoS-CoM distance of LSS1 was greater than USS and LSS2 of CSB and MSS. The larger BoS-CoM distance provides greater gravitational deceleration.

During the stepping phase, USS and LSS2 may benefit from a more favorable BoS-CoM distance than LSS1. Following LSS1 lift-off, the CoM is more lateral than the BoS resulting in gravitational acceleration in the direction of the fall. Conversely, during a portion of the stepping phase of USS and in LSS2, the BoS may be more lateral to the CoM and result in gravitational deceleration. The lower CoM velocity of CSF, MSS, and LSS2 at step loading support this possibility. Additionally, at first step loading, CSF and CSB have positive MoS and non-zero  $\tau$  implying that some deceleration may occur prior to LSS2.

Multistep responses have been suggested to be part of a pre-planned strategy to recover balance through a series of two or more postural adjustments (Luchies et al., 1994; Maki et al., 2000). Alternatively, using a multistep strategy may be an indicator of impaired weight shifting or stepping abilities. An USS-LSS2 protective stepping sequence obviates the need to rapidly shift the body weight from the passively loaded to unloaded limb prior to the first step, as is required to perform a LSS. However, rapid lateral weight shift is required in USS-LSS2 after the first step. Second step unloading began about 193 ms prior to the loading definition of the first step in USS-LSS2 trials indicating that lateral weight transfer occurs in an USS-LSS2 (after the first step). Overall, recovering balance using USS-LSS2 appears more problematic compared with LSS1 steps. In addition to increased fall risk using multiple steps (Johnson-Hilliard et al., 2010), less stable and more complex stepping trajectory and inter-limb collisions (Borrelli et al., 2019; Maki et al., 2000; Mille et al., 2013), USS-LSS2 do not avoid a rapid shift in body weight support, it is simply delayed from the 1st to the 2nd step.

Control of the CoM in the frontal plane is reliant on the hip abductor-adductor (AB-AD), trunk, and ankle musculature (Rietdyk et al., 1999; Winter et al., 1996). LSS1 requires rapid phasic neuromuscular activation of the hip AB-AD muscles for lateral weight transfer prior to step lift off (Mille et al., 2005; Inacio et al., 2019). Impaired hip AB-AD neuromuscular performance has been associated with a reduced incidence of LSS1 among a similar cohort of older adults as studied here (Inacio et al., 2019). Accumulating evidence suggests that diminished hip AB-AD capacity plays a role in the prevalence of USS-LSS2. Older adults have been shown to have decreased AB-AD performance in isolated rate of torque development assessments and in stepping assessments which was associated with a decreased frequency of lateral side-steps (Inacio et al., 2019; Mille et al., 2005). Power training improves AB-AD rate of torque development and increases the incidence of single lateral sidesteps (Inacio et al., 2019). However, without direct measurement of the hip AB-AD torques, this does not rule out other age-related changes that may affect stepping (Tricco et al., 2017). Future studies should determine the hip AB-AD torques and/or impairment associated with each step type and subsequent step.

Step length and BoS increase were largely directed in the frontal plane for LSS1 and MSS-LSS2. Although CSF-LSS2 or CSB-LSS2 stepping are primarily directed laterally, we found that out-of-plane AP BoS movements were increased by about 10% height over LSS1. This increase likely reflects efforts aimed at avoiding interlimb collision and results in more complex stepping trajectories (Maki et al., 2000; Mille et al., 2013) including a greater out-of-plane challenge to stability than LSS1. For example, CSF involves greater

anterior CoM velocity and stepping trajectory, and greater pelvic rotation and associated foot rotation compared to LSS1.

#### 4.1. Limitations

Among the limitations of the study, approximation of the CoM location as the midpoint of the bilateral superior iliac spine may affect estimates of the CoM location and MoS. Several studies have compared similar CoM approximations with various whole-body models that weigh the contribution of each body segment (e.g. Eames et al., 1999; Huntley et al., 2017; Inkol et al., 2018; Tisserand et al., 2016). However, there is no “gold-standard” body CoM estimation model, and assumptions associated with such models also affect estimates of CoM location (e.g. Catena et al., 2017). During feet-in-place AP and ML perturbations, approximating the CoM location at the sacrum resulted in up to 8 cm (about 5% height) and 2.5 cm (about 2% height) error in the AP and ML directions respectively (Inkol et al., 2018). CoM approximation significantly affected the MoS in the AP direction but not in the ML (Inkol et al., 2018). In the present study, all differences in MoS were larger than 2% height in the ML direction, with some differences being larger than 5% height the in AP direction. Therefore, significant differences in MoS in the AP direction should be interpreted with caution.

We determined step unloading onset and loading using kinematic criteria (vertical ankle velocity) similar to previous studies of lateral perturbation (e.g., Fujimoto et al., 2017; Johnson-Hilliard et al., 2010; Mille et al., 2013; Patton et al., 2006; Yungheer et al., 2012). While toe-off and heel-strike derived from force platform data has been traditionally used in studies of stepping, this is not feasible for lateral stepping from stationary standing because both feet may occupy the same force platform at first step loading which occurred in the present study. We have, however, compared the difference between this kinematic approach and step events defined by kinetics using limited available data. We found that, compared with kinetic determination, foot lift-off occurred about 150 ms after step unloading onset and foot contact occurred about 160 ms before step loading (see Appendix A1).

## 5. Summary

Compared with passively loaded limb single lateral steps (LSS1) induced by lateral waist-pulls, lateral side steps (LSS2) that follow passively unloaded limb first steps (USS) are characterized by slower CoM velocity, smaller BoS increase, decreased BoS-CoM distance, and decreased MoS (with the exception of the MoS of a LSS2 following a CSF). CSF-LSS2 and CSB-LSS2 also increased performance demand by increasing movements and stability requirements in the sagittal plane necessitating more complex stepping trajectories than for single lateral sidesteps.

## Acknowledgements

The assistance of Woei-Nan Bair, Brock Beamer, Alexandra Cirillo, Masahiro Fujimoto, Valentina Graci, Hao-Yuan Hsiao, Mario Inacio, Chen-Chieh Lin, Douglas Pizac, Kaitlin Riddle, Ozell Sanders, Douglas Savin, Kathy Simpson, John Sorkin, Tricia Young, and the LIFT study team is gratefully acknowledged. We thank Professor Marie-Laure Mille for suggestions on data analysis. The assistance of the Baltimore VA Medical Center, Geriatric Research, Education and Clinical Center is gratefully acknowledged.



## Funding

NIH/NIA grant R01AG033607, the University of Maryland Claude D Pepper-OAIC NIH/NIA grant P30AG028747, the UMANRRT Program (NIDRR 90AR50280, NIDILRR 90AR5004 formerly H133P100014).

## Appendix A

In the present study, beginning and end of stepping events were defined using vertical ankle velocity. The vertical ankle velocity-based stepping event identification method used in the present study, and another kinematic based definition are compared with the results of trials where force platform data were available. Step lift-off and contact were determined using force platforms in 148/217 and 60/217 trials. Technical problems resulted in a loss of data for 69 trials. The determination of stepping events using force platforms is complicated by the landing location of the first step and subsequent steps dependent upon step type. In many trials, the first step landed on the same platform as the stance leg making stepping events indeterminate. Second step lift-off and contact were determined in 27/155 and 25/155 trials respectively (there were 155 trials with a 2-step sequence).

### Stepping event definition

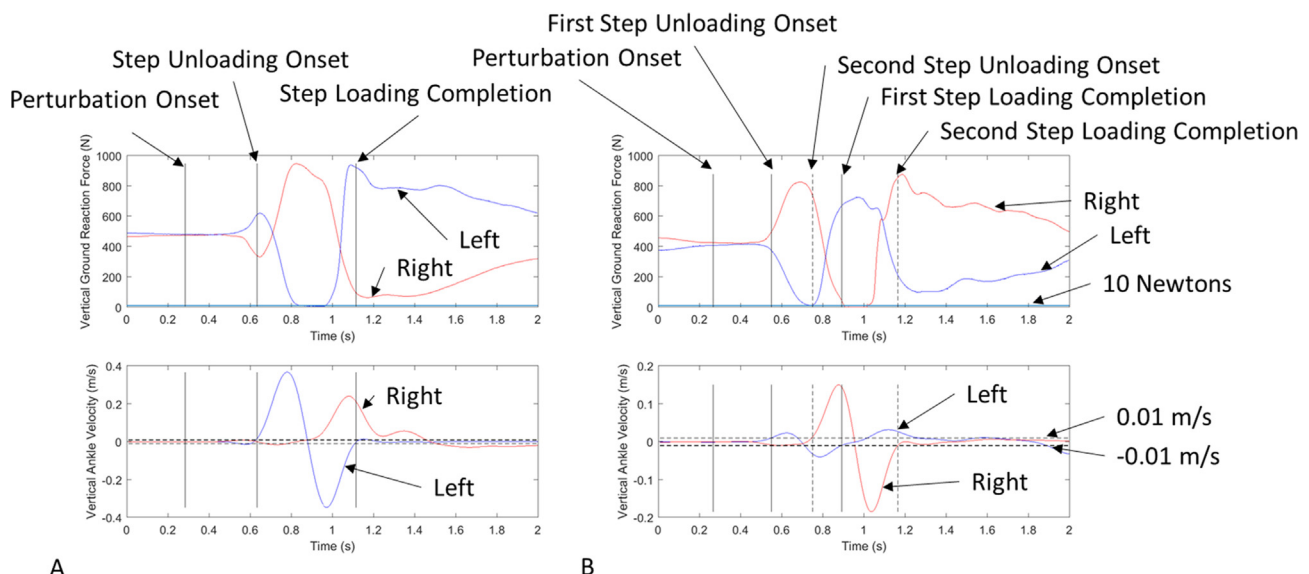
Two kinematic-based step event definitions are compared with stepping events determined using force platforms. The first definition, used in the present study, uses vertical ankle velocity. Step unloading onset is defined as the time when the vertical ankle velocity is greater than 0.01 m/s and step loading is defined as the time when the vertical ankle velocity is  $<0.01$  m/s in the downward direction. The second definition uses vertical heel and toe position to define step events. Step onset and step completion were respectively defined as the time the toe and heel markers were simultaneously more than 0.01 m above the ground or the toe or heel markers were within 0.01 m of the ground. Finally, step lift-off and touch down were defined using force platforms with the

vertical ground reaction force (vGRF) being less than 10 Newtons and greater than 10 Newtons respectively.

Fig. A1 shows the vGRF and vertical ankle velocity of the left and right limb. A single lateral sidestep (LSS1) is shown in Fig. A1A and a medial sidestep (MSS) followed by a lateral sidestep (MSS-LSS2) is shown in Fig. A1B. Table A1 summarize the vGRF at first and second step unloading onset using the vertical ankle velocity-based step event definition, the difference in timing, and the Pearson correlation of stepping events using the ankle velocity definition described above and that found using force plates. First step kinematic unloading onset occurred 120–185 ms before force platform defined step lift-off (vGRF  $< 10$  N, Table A1). Similarly, first and second step loading occurred about 167 ms after force platform contact (Table A2).

Table A3 and Table A4 summarize the results when vertical toe and heel vertical height were used to define stepping events. As a reminder, the stepping events using this algorithm are termed step initiation and step completion. The stepping limb was completely unloaded or nearly unloaded as indicated by the vGRF at first and second step onset. First and second step onset occurred about 10–20 ms after lift-off occurred according to force plate estimates (Table A3). First and second step completion occurred 6–20 ms before lift-off occurred according to force platform estimates (Table A4). The estimates of step completion were quite close, within 20 ms, to the timing of stepping events estimated by force platforms.

Our step event definition is not consistent with the traditional step events defined by toe-off and heel-strike during steady state gait. Following our definition, the unloading/loading phase includes the stepping phase and an unloading phase prior to stepping and a loading phase following stepping with the unloading and loading phases taking about 150 ms each. Several kinematic based step event definitions were considered (in addition to those presented here), and ultimately vertical ankle velocity was selected for its simplicity and the fact that it represented the point in time that the landing limb was loaded with the more than majority of the weight (78–87%BW, see Table A2) of the body regardless of whether the foot landed with the toe or heel first and/or accepted weight with a raised heel or toe. The CoP location is to a large



**Fig. A1.** Example data from a leftward pull evoking a single lateral sidestep (A) and rightward pull evoking a medial sidestep followed by a lateral sidestep (B). The vertical ground reaction force of the left (blue) and right limb (red) is shown in the top panels and the vertical ankle velocity of the left (blue) and right (red) ankle is shown in the bottom panels. Step unloading onset and step loading are defined as the time that the vertical ankle velocity is greater than 0.01 m/s in the upward and downward direction respectively. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

**Table A1**

Mean (standard deviation) vertical ground reaction force (percent body weight) at the beginning of the first and second step (termed “step unloading onset”) determined using kinematics. First and second step onset were identified using vertical ankle velocity was greater than 1 cm/s. The timing of stepping events was also determined using a force plate (vertical ground reaction force less than 10 N) and compared to the timing determined using vertical ankle velocity, the mean (standard deviation) difference in timing between kinematic and kinetic event detection is reported in milliseconds. The correlation and number of trials (*n*) used are also reported. Due to technical difficulties or both feet occupying the same force plate, the force plates (kinetics) could not be used to identify stepping events in many trials.

Step Type	Ankle Velocity Based Step Event Definition			
	First Step Unloading Onset 1		Second Step Unloading Onset	
	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle velocity and force platform-based timing events (ms)	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle velocity and force platform-based timing events (ms)
Lateral Sidestep	71 (18)	148 (48)	–	–
Cross Step to the Front	31 (12)	184 (95)	–	–
Cross Step to the Back	29 (15)	121 (99)	–	–
Medial Sidestep	33 (12)	142 (64)	91 (16)	167 (31)
Average	44 (23)	143 (72)	91 (16)	167 (31)
	Correlation	n	Correlation	n
Lateral Sidestep	0.87	45	–	0
Cross Step to the Front	0.06	12	–	0
Cross Step to the Back	0.54	29	–	0
Medial Sidestep	0.71	62	0.99	27
Average	0.68		0.99	

**Table A2**

Mean (standard deviation) vertical ground reaction force (percent body weight) at the end of the first and second step determined using kinematics. First and second step completion (termed “step loading”) were defined as the time the vertical ankle velocity was less than 1 cm/s in the downward direction. The timing of stepping events was also determined using a force plate (vertical ground reaction force greater than 10 N) and compared to the timing determined using vertical ankle velocity, the mean (standard deviation) difference in timing between kinematic and kinetic event detection is reported in milliseconds. The correlation and number of trials (*n*) used are also reported. Due to technical difficulties or both feet occupying the same force plate, the force plates (kinetics) could not be used to identify stepping events in many trials.

Step Type	Ankle Velocity Based Step Event Definition			
	First Step Loading		Second Step Loading	
	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle velocity and force platform-based timing events (ms)	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle velocity and force platform-based timing events (ms)
Lateral Sidestep	87 (26)	–153 (69)	–	–
Cross Step to the Front	–	–	–	–
Cross Step to the Back	–	–	–	–
Medial Sidestep	80 (22)	–158 (47)	78 (16)	–172 (52)
Average	83 (24)	–156 (58)	78 (16)	–172 (52)
	Correlation	n	Correlation	n
Lateral Sidestep	0.73	29	–	0
Cross Step to the Front	–	0	–	0
Cross Step to the Back	–	0	–	0
Medial Sidestep	0.95	31	0.96	25
Average	0.89		0.96	

extent a function of the percent body weight supported by each limb (Borrelli et al., 2020). At initial contact of the stepping foot ( $vGRF > 10$  N), the CoP is likely contralateral to the CoM with respect to the pull direction. Therefore, the CoM will continue to accelerate toward the BoS boundary for some time, reducing the BoS-CoM distance and increasing the CoM velocity and XCoM. As

a result, estimates of MoS at initial contact may be generous for our ‘clinical’ population of older adults who may not be capable of rapidly transporting the CoP to BoS boundary. Therefore, we have chosen to examine the MoS following limb loading when the BoS-CoM distance is likely reduced and may be a more conservative estimate of stability than at initial contact. However, future

**Table A3**

Mean (standard deviation) vertical ground reaction force (percent body weight) at first and second step onset determined using heel and toe vertical position. First and second step onset was defined as the time that the height of the toe and heel marker positions were greater than 1 cm. The timing of stepping events was also determined using a force plate (vertical ground reaction force greater than 10 N) and compared to the timing determined using vertical foot position (toe and heel elevated more than 1 cm), the mean (standard deviation) difference in timing between kinematic and kinetic event detection is reported in milliseconds. The correlation and number of trials (*n*) used are also reported. Due to technical difficulties or both feet occupying the same force plate, the force plates (kinetics) could not be used to identify stepping events in many trials.

Step Type	Foot Position Based Step Event Definition			
	First Step Onset		Second Step Onset	
	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle position and force platform-based timing events (ms)	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle position and force platform-based timing events (ms)
Lateral Sidestep	11 (19)	11 (36)	–	–
Cross Step to the Front	1 (1)	–75 (54)	–	–
Cross Step to the Back	0 (0)	–84 (64)	–	–
Medial Sidestep	2 (4)	–44 (71)	13 (17)	25 (28)
Total	5 (12)	–38 (71)	13 (17)	25 (28)
	Correlation	n	Correlation	n
Lateral Sidestep	0.92	45	–	0
Cross Step to the Front	0.73	12	–	0
Cross Step to the Back	0.88	29	–	0
Medial Sidestep	0.82	62	0.99	27
Average	0.79		0.99	

**Table A4**

Mean (standard deviation) vertical ground reaction force (percent body weight) at first and second step completion determined using heel or toe vertical position. First and second step completion was defined as the time that the toe or heel marker position was less than 1 cm. The timing of stepping events was also determined using a force plate (vertical ground reaction force greater than 10 N) and compared to the timing determined using vertical foot position (toe and heel lower than 1 cm above the ground), the mean (standard deviation) difference in timing between kinematic and kinetic event detection is reported in milliseconds. The correlation and number of trials (*n*) used are also reported. Due to technical difficulties or both feet occupying the same force plate, the force plates (kinetics) could not be used to identify stepping events in many trials.

Step Type	Foot Position Based Step Event Definition			
	First Step Completion		Second Step Completion	
	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle position and force platform-based timing events (ms)	Vertical Ground Reaction Force (Percent Body Weight)	Difference between ankle position and force platform-based timing events (ms)
Lateral Sidestep	20 (39)	0 (31)	–	–
Cross Step to the Front	–	–	–	–
Cross Step to the Back	–	–	–	–
Medial Sidestep	10 (16)	–19 (42)	6 (9)	3 (22)
Average	15 (30)	–10 (37)	6 (9)	3 (22)
	Correlation	n	Correlation	n
Lateral Sidestep	0.97	29	–	0
Cross Step to the Front	–	0	–	0
Cross Step to the Back	–	0	–	0
Medial Sidestep	0.96	31	0.99	25
Average	0.96		0.99	

work is required to determine the most appropriate time/s to evaluate the MoS during periods of lateral weight transfer.

## Appendix B

In addition to the dependent variables described in Table 1, the orientation of the foot and the pelvis in the transverse plane have

been included to further characterize differences between the stepping strategies. The change in foot angle between at step loading and step unloading onset was calculated for the stepping foot. The angle was estimated as the angle between the anteroposterior (AP) axis and the vector connecting the heel and the midpoint of the great toe and fifth metatarsophalangeal joint. Positive angles represent an external orientation and negative angles represent

internal orientation. The change in the angle of the pelvis was determined between step loading of the first or second step and step unloading onset. The angle of the pelvis was calculated as the angle between the mediolateral (ML) axis and the vector connecting the bilateral posterior superior iliac spine relative to the stepping leg at step loading. Pelvis rotation is reported as positive internal rotation about the leg ipsilateral to the pull regardless of step type or step number in an effort to facilitate comparison between step types.

### Anterior-Posterior LSS1 and USS characteristics at first step loading

First step type significantly affected the biomechanical characteristics of USS compared to a single lateral sidestep (LSS1,  $p$ -values < 0.001, Table A5). Stepping and stability characteristics associated with LSS1 and medial sidesteps (MSS) followed by a lateral sidestep (MSS-LSS2) were largely contained in the frontal plane compared to cross-over steps (CSF/CSB). Cross-over steps to the front (CSF, 7% height,  $p$  < 0.001) and cross-over steps to the back (CSB, -14% height,  $p$  < 0.001) contained a stepping trajectory component directed to the anterior and posterior respectively. Stepping did not substantially affect the AP BoS increase (<1% height) although CSB (-4% height,  $p$  < 0.001) resulted in a significant decrease in the BoS length compared to LSS1 (0% height).

In general, the BoS-CoM distance was larger for LSS1 at first step loading, particularly in the anterior direction of CSB where the CoM was 4% height outside of the BoS and significantly different than LSS1 (6% height,  $p$  < 0.001). The absolute CoM velocity at step loading was significantly larger for USS (CSF: 4% height/s,  $p$  < 0.001; CSB: -9% height,  $p$  < 0.001; MSS: -5% height,  $p$  < 0.001), compared to LSS1 (-2% height). The MoS was larger for LSS1 (anterior: 7% height; posterior: 8% height) than CSB (-1% height,  $p$  < 0.001) in the anterior direction and larger than CSF (6% height,  $p$  = 0.011) and MSS (7% height,  $p$  = 0.008) in the posterior direction.

The posterior MoS for CSF (6% height,  $p$  = 0.011) and MSS (7% height,  $p$  = 0.008) and anterior MoS for CSB (-1% height,  $p$  < 0.001) were smaller than LSS1 at first step loading (anterior: 7% height, posterior: 8% height). Step type did not significantly affect the time it will take the XCoM to reach the BoS boundary,  $\tau$ , in the anterior direction ( $p$  = 0.43). Although  $\tau$  was significantly affected in the posterior direction ( $p$  < 0.001), post-hoc comparisons failed to identify any significant differences ( $p$ -values > 0.24).

### Anterior-Posterior LSS1 and LSS2 characteristics at final step loading

The AP step length of LSS1 (0% height) was less than a LSS2 following a CSF (CSF-LSS2, -9% height,  $p$  < 0.001) and a LSS2 following a CSB (CSB-LSS2, 4% height,  $p$  < 0.001) but not significantly different from MSS-LSS2 (0% height,  $p$  = 0.90, Table A6). The AP stepping component associated with CSF-LSS2 (10% height,  $p$  < 0.001) and CSB-LSS2 (12% height,  $p$  < 0.001) significantly increased the BoS compared to LSS1 (1% height). AP CoM displacement was relatively small for LSS1, CSF-LSS2, CSB-LSS2, and MSS-LSS2 (<3% height). The anterior and posterior BoS-CoM distance was larger for CSF-LSS2 (anterior: 10% height; posterior: 14% height) and CSB-LSS2 (anterior: 12% height; posterior: 14% height) compared to LSS1 (anterior: 7% height, posterior: 9% height;  $p$ -values < 0.001).

The absolute CoM velocity at step loading was larger for CSF-LSS2 (-13% height/s) and CSB-LSS2 (6% height/s) compared to LSS1 (-1% height,  $p$ -values < 0.001). The MoS of LSS1 (anterior: 7% height; posterior: 8% height) was smaller than CSF-LSS2 (anterior: 14% height,  $p$  < 0.001; posterior: 10% height,  $p$  = 0.042) and CSB-LSS2 (anterior: 10% height,  $p$  < 0.001; posterior: 16% height,  $p$  < 0.001) and MSS-LSS2 in the posterior direction (10% height,  $p$  = 0.018). The anterior MoS of LSS1 (7% height) was significantly larger than MSS-LSS2 (6% height,  $p$  = 0.011) while the posterior MoS of LSS1 (8% height,  $p$  = 0.018) was significantly smaller than MSS-LSS2 (10% height). First step type did not have a significant effect on  $\tau$  ( $p$ -values < 0.16).

### Foot and pelvis orientation in the transverse plane

There was not a significant difference in change in the angle of the foot and pelvis (<4 degrees, Table A7) between LSS1 and MSS-LSS2 ( $p$ -values > 0.19). The foot was significantly internally and externally reoriented following CSF (-34 degrees internal) and CSB (38 degrees external) at first step loading compared to LSS1 (0 degrees,  $p$ -values < 0.001). The pelvis was significantly internally rotated and externally rotated in CSF (30 degrees internal) and CSB (-20 degrees external) at first step loading compared to LSS1 (0 degrees,  $p$ -values < 0.001). At second step loading, the foot was significantly externally and internally reoriented in CSF-LSS2 (25 degrees,  $p$  < 0.001) and CSB-LSS2 (-23 degrees,  $p$  < 0.001), respectively, compared to LSS1 (0 degrees). The pelvis was significantly internally and externally rotated for CSF-LSS2 (35 degrees,  $p$  < 0.001) and CSB-LSS2 (-26 degrees,  $p$  < 0.001) compared to LSS1 (1 degree).

**Table A5**

Estimated marginal mean (EMM) and standard error of the mean (SEM) for outcome measures at first step loading.  $P$ -values from post-hoc comparisons between a single lateral sidestep (LSS1) and cross-over step to the front (CSF), LSS1 and a cross-over step to the back (CSB), and LSS1 and a medial sidestep (MSS) are also shown. The number of trials ( $n$ ) used in calculation of the time it will take the extrapolated center of mass (XCoM) to reach the base of support,  $\tau$ , is noted under each step type. In some trials the CoM velocity was in the direction opposite of the BoS boundary and the  $\tau$  is not defined. For example, in 5/94 MSS the CoM was moving away from the posterior BoS boundary and  $\tau$  is not defined. \*Significant step type effect ( $p$  < 0.001). Negative values for step length and CoM displacement reflect movement to the posterior. There was little variation from trial to trial and the model would not converge. #A model with participant as a random effect and step type as a fixed effect was used.

Anteroposterior	LSS1	CSF		CSB		MSS	
First Step Characteristics	EMM (SEM)	EMM (SEM)	p-value	EMM (SEM)	p-value	EMM (SEM)	p-value
Step Length (% height)*	0.3 (0.4)	7.1 (0.7)	<0.001	-14.0 (0.5)	<0.001	-1.2 (0.4)	0.07
AP BoS Increase (% height)*	0.9 (0.2)	-0.9 (0.4)	<0.001	-4.0 (0.3)	<0.001	-0.9 (0.2)	<0.001
AP CoM Displacement (% height)*	-2.3 (0.2)	0.1 (0.4)	<0.001	-7.2 (0.3)	<0.001	-2.2 (0.2)	0.6
Anterior BoS-CoM Distance (% height)*	6.2 (0.5)	7.7 (0.08)	0.09	-3.5 (0.6)	<0.001	4.4 (0.4)	0.001
Posterior BoS-CoM Distance (% height)*	8.6 (0.4)	5.0 (0.6)	<0.001	13.0 (0.5)	<0.001	8.1 (0.4)	0.29
AP CoM Velocity (% height/s)*	-2.1 (0.9)	4.2 (1.4)	<0.001	-9.2 (1.0)	<0.001	-5.4 (0.8)	<0.001
Anterior MoS (% height)*	6.7 (0.6)	6.5 (0.9)	0.83	-0.6 (0.7)	<0.001	6.0 (0.5)	0.31
Posterior MoS (% height)*	8.1 (0.5)	5.9 (0.8)	0.011	10.2 (0.6)	0.004	6.6 (0.5)	0.008
Anterior $\tau$ (ms) #	9998 (5927)	1083(6879)	-	-	-	2173 (13482)	-
n	16/62	12/20		0/41		3/94	
Posterior $\tau$ (ms)#,*	5839 (1656)	9471 (4934)	0.49	3126 (1863)	0.28	3451 (1196)	0.24
n	46/62	5/20		36/41		89/94	



**Table A6**

Estimated marginal mean (EMM) and standard error of the mean (SEM) at step unloading onset and various dependent variables at first step loading. Dependent variables were calculated at the first step loading of the first step for single lateral sidesteps (LSS1) and at the step loading of the lateral sidestep (LSS2) following a cross-over step to the back (CSF-LSS2), cross-over step to the front (CSF-LSS2), and medial sidestep (MSS-LSS2). *P*-values from post-hoc comparisons between LSS1 and CSF-LSS2, LSS and CSB-LSS2, LSS and MSS-LSS2. The number of trials (*n*) used in calculation of the time it will take the extrapolated center of mass (XCoM) available to reach the base of support,  $\tau$ , is noted under each step type. In some trials the CoM velocity was in the direction opposite of the BoS boundary and the  $\tau$  is not defined. For example, in 6/41 CSB-LSS2 the CoM was moving away from the anterior BoS boundary and  $\tau$  is not defined. \*Significant step type effect ( $p < 0.001$ ). Negative values for step length and CoM displacement reflect movement to the posterior. There was little variation from trial to trial and the model would not converge. #A model with participant as a random effect and step type as a fixed effect was used.

Anteroposterior	LSS1	CSF-LSS2		CSB-LSS2		MSS-LSS2	
LSS1 and LSS2 Characteristics	EMM (SEM)	EMM (SEM)	p-value	EMM (SEM)	p-value	EMM (SEM)	p-value
Step Length (% height)*	−0.1 (0.6)	−8.5 (1.0)	<b>&lt;0.001</b>	3.5 (0.7)	<b>&lt;0.001</b>	0.02 (0.5)	0.90
AP BoS Increase (% height)*	1.3 (0.5)	9.8 (0.9)	<b>&lt;0.001</b>	12.2 (0.6)	<b>&lt;0.001</b>	1.8 (0.5)	0.32
AP CoM Displacement (% height)*	−2.1 (0.4)	−2.7 (0.6)	0.4	−2.3 (0.4)	0.72	−1.2 (0.3)	<b>0.019</b>
Anterior BoS-CoM Distance (% height)*	6.6 (0.4)	9.5 (0.7)	<b>&lt;0.001</b>	11.5 (0.5)	<b>&lt;0.001</b>	6.0 (0.4)	0.18
Posterior BoS-CoM Distance (% height)*	8.5 (0.5)	13.8 (0.7)	<b>&lt;0.001</b>	14.2 (0.6)	<b>&lt;0.001</b>	9.4 (0.4)	0.07
AP CoM Velocity (% height/s)*	−1.3 (0.9)	−12.8 (1.4)	<b>&lt;0.001</b>	6.0 (1.0)	<b>&lt;0.001</b>	1.2 (0.7)	<b>0.014</b>
Anterior MoS (% height)*	7.0 (0.5)	13.8 (0.8)	<b>&lt;0.001</b>	10.0 (0.6)	<b>&lt;0.001</b>	5.5 (0.4)	<b>0.011</b>
Posterior MoS (% height)*	8.2 (0.5)	10.0 (0.8)	<b>0.042</b>	15.8 (0.6)	<b>&lt;0.001</b>	9.6 (0.5)	<b>0.018</b>
Anterior $\tau$ (ms)#	5883 (6436)	9591 (9690)	–	10,394 (47031)	–	6337 (3877)	–
N	16/62	6/20		35/41		59/94	
Posterior $\tau$ (ms)#	20,158 (14311)	16,023 (17722)	–	21,338 (23419)	–	27,720 (14375)	–
n	46/62	14/20		6/41		35/94	

**Table A7**

Estimated marginal mean (EMM) and standard error of the mean (SEM) for foot and pelvis angles in the transverse plane. The change in foot angle is calculated between step loading and step unloading onset. The change in pelvis angle is calculated between first or second step loading and first step unloading onset. The limb ipsilateral to the pull direction is assumed to be the stance limb regardless of step type or step count in an effort to facilitate comparison of pelvic angle across step types. External rotation is positive for the feet and pelvis. \*Significant step type effect ( $p < 0.001$ ). There was little variation from trial to trial and the model would not converge. #A model with participant as a random effect and step type as a fixed effect was used.

	LSS1	CSF-LSS2		CSB-LSS2		MSS-LSS2	
	EMM (SEM)	EMM (SEM)	p-value	EMM (SEM)	p-value	EMM (SEM)	p-value
Change in horizontal plane foot angle at first step loading (degrees)#	0.5 (2.4)	−33.8 (4.1)	<b>&lt;0.001</b>	37.9 (2.9)	<b>&lt;0.001</b>	3.8 (2.0)	0.27
Pelvis rotation at first step loading (degrees)	0.3 (1.3)	30.1 (2.0)	<b>&lt;0.001</b>	−20.3 (1.5)	<b>&lt;0.001</b>	0.5 (1.1)	0.87
Change in horizontal plane foot angle at second step loading (degrees)	0.2 (2.9)	25.3 (4.8)	<b>&lt;0.001</b>	−22.5 (3.2)	<b>&lt;0.001</b>	−1.7 (2.3)	0.54
Pelvis Rotation at second step loading (degrees)	1.2 (1.6)	35.2 (2.6)	<b>&lt;0.001</b>	−25.5 (1.8)	<b>&lt;0.001</b>	−1.1 (1.3)	0.19

## References

- Bair, W.N., Prettyman, M.G., Beamer, B.A., Rogers, M.W., 2016. Kinematic and behavioral analyses of protective stepping strategies and risk for falls among community living older adults. *Clin. Biomech.* 36, 74–82. <https://doi.org/10.1016/j.clinbiomech.2016.04.015>.
- Borrelli, J., Creath, R.A., Pizac, D., Hsiao, H., Sanders, O.P., Rogers, M.W., 2019. Perturbation-evoked lateral steps in older adults: Why take two steps when one will do? *Clin. Biomech.* 63, 41–47. <https://doi.org/10.1016/j.clinbiomech.2019.02.014>.
- Borrelli, J., Komisar, V., Novak, A.C., Maki, B.E., King, E.C., 2020. Extending the center of pressure to incorporate handhold forces: Derivation and sample application. *J. Biomech.* 104, 109727. <https://doi.org/10.1016/j.jbiomech.2020.109727>.
- Catena, R.D., Chen, S.H., Chou, L.S., 2017. Does the anthropometric model influence whole-body center of mass calculations in gait? *J. Biomech.* 59, 23–28. <https://doi.org/10.1016/j.jbiomech.2017.05.007>.
- Eames, M.H.A., Cosgrove, A., Baker, R., 1999. Comparing methods of estimating the total body centre of mass in three-dimensions in normal and pathological gait. *Hum. Mov. Sci.* 18, 637–646. [https://doi.org/10.1016/S0167-9457\(99\)00022-6](https://doi.org/10.1016/S0167-9457(99)00022-6).
- Folstein, M.F., Folstein, S.E., McHugh, P.R., 1975. "Mini-Mental State" A practical method for grading the cognitive state of patients for the clinician. *J. Psychiatr. Res.* 12, 189–198.
- Fujimoto, M., Bair, W.N., Rogers, M.W., 2017. Single and multiple step balance recovery responses can be different at first step lift-off following lateral waist-pull perturbations in older adults. *J. Biomech.* 55, 41–47. <https://doi.org/10.1016/j.jbiomech.2017.02.014>.
- Hof, A.L., 2007. The equations of motion for a standing human reveal three mechanisms for balance. *J. Biomech.* 40, 451–457. <https://doi.org/10.1016/j.jbiomech.2005.12.016>.
- Hof, A.L., Gazendam, M.G.J., Sinke, W.E., 2005. The condition for dynamic stability. *J. Biomech.* 38, 1–8. <https://doi.org/10.1016/j.jbiomech.2004.03.025>.
- Huntley, A.H., Schinkel-Ivy, A., Aquil, A., Mansfield, A., 2017. Validation of simplified centre of mass models during gait in individuals with chronic stroke. *Clin. Biomech.* 48, 97–102. <https://doi.org/10.1016/j.clinbiomech.2017.07.015>.
- Hurt, C.P., Rosenblatt, N.J., Grabiner, M.D., 2011. Form of the compensatory stepping response to repeated laterally directed postural disturbances. *Exp. Brain Res.* 214, 557–566. <https://doi.org/10.1007/s00221-011-2854-1>.
- Inacio, M., Creath, R., Rogers, M.W., 2019. Effects of aging on hip abductor-adductor neuromuscular and mechanical performance during the weight transfer phase of lateral protective stepping. *J. Biomech.* 82, 244–250. <https://doi.org/10.1016/j.jbiomech.2018.10.040>.
- Inacio, M., Creath, R., Rogers, M.W., 2018. Low-dose hip abductor-adductor power training improves neuromechanical weight-transfer control during lateral balance recovery in older adults. *Clin. Biomech.* 60, 127–133. <https://doi.org/10.1016/j.clinbiomech.2018.10.018>.
- Inkol, K.A., Huntley, A.H., Vallis, L.A., 2018. Modeling margin of stability with feet in place following a postural perturbation: Effect of altered anthropometric models for estimated extrapolated centre of mass. *Gait Posture* 62, 434–439. <https://doi.org/10.1016/j.gaitpost.2018.03.050>.
- Johnson-Hilliard, M., Martinez, K.M., Janssen, I., Edwards, B.J., Mille, M.L., Zhang, Y., Rogers, M.W., 2010. Lateral balance factors predict future falls in community-living older adults. *Arch. Phys. Med. Rehabil.* 89, 1708–1713. <https://doi.org/10.1016/j.apmr.2008.01.023>.
- Luchies, C.W., Alexander, N.B., Schultz, A.B., Ashton-Miller, J., 1994. Stepping responses of young and old adults to postural disturbances: kinematics. *J. Am. Geriatr. Soc.* 42, 506–512. <https://doi.org/10.1111/j.1532-5415.1994.tb04972.x>.
- Maki, B.E., Edmondstone, M.A., McIlroy, W.E., 2000. Age-related differences in laterally directed compensatory stepping behavior. *J. Gerontol. Ser. A Biol. Sci. Med. Sci.* 55, 270–277. <https://doi.org/10.1093/gerona/55.5.M270>.
- Maki, B.E., McIlroy, W.E., Perry, S.D., 1996. Influence of lateral destabilization on compensatory stepping responses. *J. Biomech.* 29, 343–353. [https://doi.org/10.1016/0021-9290\(95\)00053-4](https://doi.org/10.1016/0021-9290(95)00053-4).
- Mille, M.L., Johnson-Hilliard, M., Martinez, K.M., Zhang, Y., Edwards, B.J., Rogers, M.W., 2013. One step, two steps, three steps more... directional vulnerability to falls in community-dwelling older people. *J. Gerontol. Ser. A Biol. Sci. Med. Sci.* 68, 1540–1548. <https://doi.org/10.1093/gerona/glt062>.
- Mille, M.L., Johnson, M.E., Martinez, K.M., Rogers, M.W., 2005. Age-dependent differences in lateral balance recovery through protective stepping. *Clin. Biomech.* 20, 607–616. <https://doi.org/10.1016/j.clinbiomech.2005.03.004>.
- Mille, M.L., Rogers, M.W., Martinez, K., Hedman, L.D., Johnson, M.E., Lord, S.R., Fitzpatrick, R.C., 2003. Thresholds for inducing protective stepping responses to external perturbations of human standing. *J. Neurophysiol.* 90, 666–674. <https://doi.org/10.1152/jn.00974.2002>.
- Nevitt, M.C., Cummings, S.R., 1993. Type of fall and risk of hip and wrist fractures: The study of osteoporotic fractures. *J. Am. Geriatr. Soc.* 41, 1226–1234.
- Pai, Y.C., Patton, J., 1997. Center of mass velocity-position predictions for balance control. *J. Biomech.* 30, 347–354. [https://doi.org/10.1016/S0021-9290\(96\)00165-0](https://doi.org/10.1016/S0021-9290(96)00165-0).

- Pai, Y.C., Rogers, M.W., Patton, J., Cain, T.D., Hanke, T.A., 1998. Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults. *J. Biomech.* 31, 1111–1118. [https://doi.org/10.1016/S0021-9290\(98\)00124-9](https://doi.org/10.1016/S0021-9290(98)00124-9).
- Patton, J.L., Hilliard, M.J., Martinez, K., Mille, M., Rogers, M.W., 2006. A Simple Model of Stability Limits Applied to Sidestepping in Young, Elderly and Elderly Fallers 3305–3308.
- Pidcoe, P.E., Rogers, M.W., 1998. A closed-loop stepper motor waist-pull system for inducing protective stepping in humans. *J. Biomech.* 31, 377–381. [https://doi.org/10.1016/S0021-9290\(98\)00017-7](https://doi.org/10.1016/S0021-9290(98)00017-7).
- Rietdyk, S., Patla, A.E., Winter, D.A., Ishac, M.G., Little, C.E., 1999. Balance recovery from medio-lateral perturbations of the upper body during standing. *J. Biomech.* 32, 1149–1158. [https://doi.org/10.1016/S0021-9290\(99\)00116-5](https://doi.org/10.1016/S0021-9290(99)00116-5).
- Rogers, M.W., Mille, M.L., 2003. Lateral stability and falls in older people. *Exerc. Sport Sci. Rev.* 31, 182–187. <https://doi.org/10.1097/00003677-200310000-00005>.
- Sawyer Radloff, L., 1977. The CES-D scale: a self-report depression scale for research in the general population. *Appl. Psychol. Meas.* 1, 385–401.
- Sturnieks, D.L., Menant, J., Delbaere, K., Vanrenterghem, J., Rogers, M.W., Fitzpatrick, R.C., Lord, S.R., 2013. Force-controlled balance perturbations associated with falls in older people: a prospective cohort study. *PLoS ONE* 8, 1–6. <https://doi.org/10.1371/journal.pone.0070981>.
- Sturnieks, D.L., Menant, J., Vanrenterghem, J., Delbaere, K., Fitzpatrick, R.C., Lord, S.R., 2012. Sensorimotor and neuropsychological correlates of force perturbations that induce stepping in older adults. *Gait Posture* 36, 356–360. <https://doi.org/10.1016/j.gaitpost.2012.03.007>.
- Tisserand, R., Robert, T., Dumas, R., Chèze, L., 2016. A simplified marker set to define the center of mass for stability analysis in dynamic situations. *Gait Posture* 48, 64–67. <https://doi.org/10.1016/j.gaitpost.2016.04.032>.
- Tricco, A.C., Thomas, S.M., Veroniki, A.A., Hamid, J.S., Cogo, E., Striffler, L., Khan, P.A., Robson, R., Sibley, K.M., MacDonald, H., Riva, J.J., Thavorn, K., Wilson, C., Holroyd-Leduc, J., Kerr, G.D., Feldman, F., Majumdar, S.R., Jaglal, S.B., Hui, W., Straus, S.E., 2017. Comparisons of interventions for preventing falls in older adults: A systematic review and meta-analysis. *JAMA - J. Am. Med. Assoc.* 318, 1687–1699. <https://doi.org/10.1001/jama.2017.15006>.
- Winter, D.A., Prince, F., Frank, J.I.M.S., Powell, C., Zabjek, K.F., 1996. Unified theory regarding A/P and M/L balance in quiet stance. *J. Neurophysiol.* 75, 2334–2343.
- Yang, F., Pai, Y.-C., 2014. Can sacral marker approximate center of mass during gait and slip-fall recovery among community-dwelling older adults?. *J. Biomech.* 47, 3807–3812. <https://doi.org/10.1016/j.jbiomech.2014.10.027>.
- Yungher, D.A., Morgia, J., Bair, W., Inacio, M., 2012. Short-term changes in protective stepping for lateral balance recovery in older adults 27, 151–157. <https://doi.org/10.1016/j.clinbiomech.2011.09.003>. Short-term.