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Kinematic and behavioral analyses of protective stepping strategies and risk for falls among community living older adults

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Abstract

Background—Protective stepping evoked by externally applied lateral perturbations reveals balance deficits underlying falls. However, a lack of comprehensive information about the control of different stepping strategies in relation to the magnitude of perturbation limits understanding of balance control in relation to age and fall status. The aim of this study was to investigate different protective stepping strategies and their kinematic and behavioral control characteristics in response to different magnitudes of lateral waist-pulls between older fallers and non-fallers.

Methods—Fifty-two community-dwelling older adults (16 fallers) reacted naturally to maintain balance in response to five magnitudes of lateral waist-pulls. The balance tolerance limit (BTL, waist-pull magnitude where protective steps transitioned from single to multiple steps), first step control characteristics (stepping frequency and counts, spatial-temporal kinematic, and trunk position at landing) of four naturally selected protective step types were compared between fallers and non-fallers at- and above-BTL.

Findings—Fallers took medial-steps most frequently while non-fallers most often took crossover-back-steps. Only non-fallers varied their step count and first step control parameters by step type at the instants of step initiation (onset time) and termination (trunk position), while both groups modulated step execution parameters (single stance duration and step length) by step type. Group differences were generally better demonstrated above-BTL.

Interpretation—Fallers primarily used a biomechanically less effective medial-stepping strategy that may be partially explained by reduced somato-sensation. Fallers did not modulate their step parameters by step type at first step initiation and termination, instances particularly vulnerable to instability, reflecting their limitations in balance control during protective stepping.

Keywords

Falls; Balance; Postural perturbation; Protective stepping

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1. Introduction

Falls among older adults are a multi-factorial problem (Tinetti et al., 1988) where neuromuscular (Tinetti et al., 1988; Guralnik et al., 1994; Hilliard et al., 2008) and sensorimotor impairments (DeMott et al., 2007; Lord et al., 2010; Inacio et al., 2014) underlying balance and gait deficits represent significant risk factors. These balance and mobility impairments underlying falls may be better understood in a dynamic and complex context by examining older adults' responses to externally applied, unpredictable postural perturbations that simulate the loss of balance leading to naturally occurring falls (Hilliard et al., 2008; Mansfield and Maki, 2009; Mille et al., 2005; Maki et al., 1996, 2000). Understanding protective stepping response to laterally oriented perturbations is of clinical importance because lateral balance is particularly challenging (Mansfield and Maki, 2009; Mille et al., 2013) for older adults who have greater fall risk (Hilliard et al., 2008) or a history of falls (Mille et al., 2005, 2013). Moreover, video surveillance of naturally occurring falls in older adults detected particular problems in controlling lateral balance during sideways falls (Holliday et al., 1990), and hip fractures occur most frequently in association with lateral falls (Greenspan et al., 1998).

Lateral challenges to standing balance involve unique biomechanical features wherein the body's center-of-mass (CoM) is initially moved passively relative to the base-of-support (BoS) such that the leg opposite to the direction of imposed CoM movement is passively unloaded (Mille et al., 2005; Maki et al., 1996, 2000). When protective stepping is used to maintain balance, this passive unloading assists with active weight transfer and permits a faster foot-lift-off with the unloaded leg (Mille et al., 2005; Maki et al., 1996, 2000; Yungher et al., 2012). It has been hypothesized that individuals with poorer balance will use unloaded-leg-stepping more frequently than loaded-leg-stepping, however, this hypothesis has not been consistently supported (Mille et al., 2005, 2013). While fallers commonly use multiple steps for balance recovery rather than a single step (Hilliard et al., 2008; Mansfield and Maki, 2009; Maki et al., 2000; Mille et al., 2013), there are conflicting reports detailing the different stepping strategies taken after lateral perturbations. This issue is important to resolve because the form of stepping used reflects the complex neuromotor control involved in responding to lateral perturbations leading to more or less effective solutions for stabilizing balance (Mille et al., 2005, 2013; Yungher et al., 2012).

In addition to loaded limb lateral steps, unloaded-leg-stepping may involve different subtypes including crossover-steps (Mille et al., 2005; Maki et al., 1996, 2000; Hurt et al., 2011) (either frontward or backward (Yungher et al., 2012)), and medial-first steps taken either alone (Yungher et al., 2012) or as a part of an inter-limb medial-lateral side-step-sequence (Mille et al., 2005; Maki et al., 1996, 2000; Yungher et al., 2012; Hurt et al., 2011). Furthermore, reports on the different subtypes have been limited for various reasons including analysis of only steps where the BoS was extended beneath the falling CoM (Maki et al., 2000), not reporting medial-steps because of their low frequency (Mille et al., 2005), or reporting only crossover-steps because of their more common occurrence (Hilliard et al., 2008; Yungher et al., 2012). Such restricted information comparing the different types of stepping limits fuller understanding of the neuromotor control strategies for maintaining balance stability and preventing falls with aging.

One influential factor affecting laterally-evoked stepping is the magnitude of perturbation (Maki et al., 1996; Meyer et al., 2004). Prior studies either used a single perturbation magnitude that always induced stepping (Hilliard et al., 2008; Mille et al., 2005, 2013), or included a systematic range of perturbation magnitudes that rarely, usually, or frequently evoked stepping without detailed characterization of the spatio-temporal parameters of each stepping type in relation to magnitude (Maki et al., 1996; Yungher et al., 2012; Meyer et al., 2004). Therefore, it is not clear how perturbation magnitude may influence the use of different stepping strategies and the identification of performance differences among fallers and non-fallers.

To further address the foregoing issues, we applied lateral waist-pull perturbations of standing balance at different magnitudes to older adults with and without a history of falls. The aims of the study were to: 1) Compare between groups the waist-pull magnitude where protective stepping transitions from single to multiple balance recovery steps (this waist-pull magnitude is defined as the Balance Tolerance Limit, BTL); 2) investigate if stepping frequency and number of steps taken differ in relation to group, step type and BTL; and 3) examine if first step spatio-temporal kinematic parameters differ in relation to group, step type and BTL. We hypothesized that the BTL would occur at a lower perturbation magnitude for the fallers compared with non-fallers, and that the stepping types used and their associated control characteristics would differ in relation to the perturbation magnitude and fall status.

2. Method

2.1. Participants

Community dwelling adults over 65-years-old were recruited from the greater Baltimore area, and from the GRECC of the Baltimore VA Medical Center. Volunteers were first screened by phone and then if qualified were medically examined by a geriatrician. Exclusion criteria were: 1) Mini Mental State Examination ≤ 24 ; 2) centers for Epidemiological Studies Survey ≤ 16 ; 3) sedative use; 4) any clinically significant functional impairment related to musculoskeletal, neurological, cardiopulmonary, metabolic, or other general medical problems that limit functional activities; 5) non-ambulatory or use of walking device at home; 6) participating in vigorous exercises or muscle strengthening exercises; 7) advised not to exercise by primary care physician; and 8) received surgery in the past year. Participants gave written informed consent according to procedures approved by the IRB of University of Maryland, and Baltimore and VA Medical Center. Subjects visited the testing laboratory once for about 3 h. Participants were divided into faller and non-faller groups based on their self-reported fall history in the past year. Testers were blinded to participants' fall history during testing. A total of 52 participants were reported with 16 fallers (mean 73.4 (standard deviation 4.6) years-old, 10 females) and 36 non-fallers (74.6 (7.6) years-old, 17 females).

2.2. Testing procedures, instructions and protocol

Participants were given verbal explanations but no physical demonstration of the waist-pulls. They were instructed to respond naturally and prevent themselves from falling. Participants

wore their normal walking shoes and stood quietly using an individually standardized stance-width based on their anthropometrics (i.e., shoulder width). This standardization method is similar to that used in previous studies (i.e., 11% (Maki et al., 2000) or 20% (Mille et al., 2005) height) to minimize potential impact of stance variation on recovery step recovery responses. Participants stood with each foot on a separate force platform (AMTI, Newton, MA, USA) as the starting position before the onset of each waist-pull. The foot locations were traced onto the platform surface to ensure consistent initial foot placement across trials for each participant. The ground reaction forces were recorded at 600 Hz. Reflective markers were placed according to Eames et al. (1999) and kinematics were recorded by a six-camera Vicon motion analysis system (Vicon 460, Oxford, UK) at 120 Hz for 7 s for each trial. Participants wore a safety harness that rescued them if they fell but did not otherwise restrict their movement. The harness system is designed to move with the participants in the frontal plane as they took recovery steps. Most participants regained balance by taking steps before the harness reached its travel limit. Only a few participants (1 faller and 3 non-fallers, each with 1 trial at the beginning of testing) were caught by the harness before they regained balance. These caught-by-harness trials were excluded from analysis. An inelastic adjustable belt, snugly secured around the waist, was aligned in participants' frontal plane at pelvis level so that the waist-pulls were applied in the medio-lateral directions. Subjects held a light cylinder in front of their body at waist level to prevent blocking the hip markers before the onset of waist-pulls. They were allowed to do anything with the rod after waist-pull onset.

Lateral waist-pulls were applied by a position-controlled motor-driven system (Pidcoe and Rogers, 1998) at five different magnitudes (from smallest magnitude-1 to largest magnitude-5; displacement: 4.5–22.5 cm, velocity: 8.6–50.0 cm/s, acceleration: 180–900 cm/s²) in both the right and left directions.

Selection of the waist-pull magnitude values was based on our previous studies of stepping responses in older and younger adults where we identified a parametric range of displacement–velocity–acceleration combinations that produced perturbations where steps were reliably less likely to occur (levels 1–2), likely to occur (level 3) and always occurred (levels 4–5) with or without multiple steps. This waist-pull magnitude range has been used to identify the threshold differences in taking protective steps between young and older adults (Mille et al., 2003) and to demonstrate short-term adaptive changes in stepping behavior, including multiple-steps, commonly observed in fallers (Hilliard et al., 2008; Mansfield and Maki, 2009; Maki et al., 2000; Mille et al., 2013; Yungheer et al., 2012). Six trials were administered for each magnitude and for each direction with a total of 60 waist-pulls. Trials were pseudo-randomly arranged by block (5 magnitudes $\times$ 2 directions in each block).

2.3. Data analysis

Step count was first determined for each balance recovery trial. Then, for each participant, stepping frequency (number of trials with stepping response), mean step count (total step count divided by the number of waist-pulls) and the Balance Tolerance Limit (BTL, the lowest waist-pull magnitude at which the mean step count was greater than one) were

determined. Perturbation magnitude effects on stepping responses were analyzed with reference to BTL (e.g., for a participant with BTL at magnitude-2, magnitude-1 responses were below-, magnitude-2 responses were at-, and responses from magnitude-3 to magnitude-5 were above-BTL). This procedure effectively normalizes perturbation magnitudes according to where participants transitioned to multiple stepping behavior (Yungher et al., 2012). For stepping trials, the first step type was categorized as either a lateral-side-step (LSS) where the passively loaded leg moves sideways in the waist-pull direction, crossover-front-step (CFS) and crossover-back-step (CBS) whereby the passively unloaded leg moves toward and past the loaded leg in front or behind the body, or medial-step (MS) where the unloaded leg moves toward but not past the loaded leg (Fig. 1) (Yungher et al., 2012). Inter-limb collisions were observed only for unloaded leg stepping. There was no obvious difference of the % of trials with collisions between fallers and non-fallers (fallers: 4.6%, non-fallers: 3.7%), and 80.7% of the collision responses were associated with multiple steps as previously reported (Maki et al., 2000). Spatio-temporal kinematic parameters of the first step were determined for onset time (when the stepping leg ankle marker first moved upward, and was also validated by the vertical ground reaction force recording) as previously reported (Mille et al., 2005), single stance duration (time between stepping leg zero vertical ground force and landing), step length (in the medio-lateral direction, expressed as a percentage of body height), and trunk angular position in the 3D space relative to the gravity line at first step landing (gravity line passing through the pelvic center calculated from four pelvic markers, and trunk alignment as the line connecting the pelvic center and the midpoint between two acromion markers). Unlike previous studies with single perturbation magnitude that examined trunk movement from perturbation onset to first step landing (Mille et al., 2005; Hurt et al., 2011), we examined trunk position only at first step landing because our protocol implemented multiple perturbation magnitudes that produce different mechanical effects on the initial trunk response making it not representative of participants' active control. Unlike previous studies that examined trunk position in the lateral direction (Mille et al., 2005; Hurt et al., 2011), we examined trunk position in the 3D space as a proxy of the destabilization effect generated by trunk which is not limited to the lateral direction at first step landing (e.g., more destabilization in the antero-posterior direction for crossover-front-step and crossover-back-step).

2.4. Statistical analyses

SAS statistical software (version 9.2) was used for data analyses. Between-group difference in BTL (Table 1) was compared using Wilcoxon–Mann–Whitney exact test. For stepping frequency (Table 2), main group and main perturbation magnitude effects were first examined. Then at- and above-BTL (below-BTL not examined due to low incidence of ~10% all stepping trials), we tested if stepping frequency depended on both group and step type, followed by post-hoc analyses of simple group effect (if stepping frequency differed between groups for a given step type) and simple step type effect (if stepping frequency differed by step type for a given group). Mean step count and first step spatio-temporal kinematic parameters were each analyzed by a separate mixed repeated measures ANOVA model with group as the between-subject factor, and step type and perturbation magnitude (at- and above-BTL only) as the repeated within-subject factors. When certain step types did

not occur for a specific participant, response variables for those step types were treated as missing data. Model specifications were compound symmetry covariance structure, unequal group variance, and Kenward–Roger adjustment. Each mixed ANOVA model was first tested for main effects (i.e., main group, step type and perturbation magnitude), and then tested for group by step type interaction at- and above-BTL. Post-hoc analyses of simple group effect (if spatio-temporal parameters differed between groups for a given step type) and simple step type effect (if spatial–temporal parameters differed by step type for a given group) were performed. Pair-wise comparisons between any two step types (total 6 comparisons) were implemented with Bonferroni adjustment to control for family-wise type I error and adjusted p values were reported accordingly. A significance level was set at p or adjusted p < 0.05.

3. Results

3.1. BTL

The numbers of participants where the BTL occurred for each waist-pull magnitude are presented in Table 1. There was no significant group difference (Wilcoxon two sample exact test, $p = 0.221$). It was noteworthy that for the highest BLT levels observed, BTL 3 and BTL 4, 78% of the non-fallers compared with 56% of fallers began to use multiple steps for balance recovery.

3.2. Stepping frequency

Overall, fallers had a higher stepping frequency than non-fallers ($\chi^2 = 9.56$, $df = 1$, $p = 0.002$; fallers: 74.3% of trials, non-fallers: 68.5%) and frequency increased with higher perturbation magnitude ($\chi^2 = 1955.54$, $df = 2$, $p < 0.0001$; above-BTL: 100%, at-BTL: 95.3%, below-BTL: 18.9%). Stepping frequency depended on both group and step type above-BTL ($\chi^2 = 47.47$, $df = 3$, $p < 0.0001$) and at-BTL ($\chi^2 = 11.41$, $df = 3$, $p = 0.0097$). Table 2 shows the results from post-hoc analyses where fallers used medial-steps (MS) most frequently, and non-fallers most often took crossover-back-steps (CBS). Group differences in stepping frequency were significant for every step type above-BTL, but at-BTL only for MS and CBS, indicating that differences in stepping frequency between fallers and non-fallers were better demonstrated above-BTL. Note that the above-described step type frequency is averaged over the entire test session and for each group. Individually, participants demonstrated several step types (Fallers: 4 types/7 subjects, 3 types/17 subjects, 2 types/9 subjects and 1 type/3 subjects; Non-fallers: 4 types/2 subjects, 3 types/10 subjects, and 2 types/4 subjects). We did not observe that the same individual consistently used one step type for below-, and another step type for above-BTL responses.

3.3. Mean step count

Overall mean step count was greater in fallers ($F_{1,19.2} = 9.87$, $p = 0.0053$; for fallers: mean 2.64 (standard error 0.22), non-fallers: 1.92 (0.08) steps), increased with perturbation magnitudes ($F_{1,108} = 57.58$, $p < 0.0001$; above-BTL: 2.66 (0.12), at-BTL: 1.91 (0.13) steps), and varied by step type ($F_{3,111} = 5.29$, $p = 0.0019$; LSS: 2.12 (0.16), CFS: 2.17 (0.16), CBS: 2.25 (0.14), and MS: 2.59 (0.13) steps) with MS having the highest step count. Group by step type interactions were significant above-BTL ($F_{7,114} = 4.39$, $p = 0.0002$) and at-BTL

($F_{7,115} = 2.60$, $p = 0.0159$). Fig. 2 shows the results from post-hoc analyses indicating that non-fallers varied their step count by step type (see bracketed comparisons) but fallers did not. For each step type, group differences (i.e., comparing two adjacent bars for the same step type) were significant for two step types (LSS, and CFS) above-BTL, but at-BTL only for LSS, indicating that groups were better differentiated above-BTL.

3.4. First step onset time

Overall, first step onset time was not different between groups ($F_{1,20.2} = 1.07$, $p = 0.3135$; fallers: 189.3 (22.3), non-fallers: 154.0 (10.0) milliseconds), but became slower with larger perturbation magnitude ($F_{1,70.5} = 5.18$, $p = 0.0259$; above-BTL: 182.5 (10.5), at-BTL: 140.2 (10.3) milliseconds), and varied by step type ($F_{3,66} = 3.88$, $p = 0.0129$; LSS: 189.4 (17.1), CFS: 158.2 (17.3), CBS: 187.2 (16.1), and MS: 151.7 (13.8) milliseconds) with MS having the fastest onset timing. Group by step type interactions were significant for above-BTL ($F_{7,79.9} = 2.17$, $p = 0.0461$) and at-BTL ($F_{6,88.6} = 2.26$, $p = 0.0449$). Fig. 3 shows the results from post-hoc analyses whereby non-fallers varied their first step onset time by step type (see bracket comparisons) but fallers did not. Generally, step onset time does not differ between fallers and non-fallers for any step type except for MS at-BTL. In summary, group step onset timing differences mainly resided in fallers' lack of modulating first step onset time according to the type of step.

3.5. First step single stance duration

Overall first step single stance duration was not different between groups ($F_{1,35.4} = 0.26$, $p = 0.6113$; fallers: 471.8 (20.4), non-fallers: 485.5 (17.5) milliseconds), but became shorter with larger perturbation magnitude ($F_{1,179} = 39.69$, $p < 0.0001$; above-BTL: 411.0 (16.9), at-BTL: 546.3 (17.5) milliseconds), and varied by step type ($F_{3,200} = 77.80$, $p < 0.0001$; LSS: 290.9 (27.0), CFS: 661.0 (25.4), CBS: 630.5 (20.2), and MS: 332.2 (20.1) milliseconds) with LSS having the shortest duration. Group by step type interactions were significant above-BTL ($F_{7,136} = 16.28$, $p < 0.0001$) and at-BTL ($F_{7,137} = 24.42$, $p < 0.0001$). Fig. 4 shows the results from post-hoc analyses that both fallers and non-fallers (see bracketed comparisons for each group) adjusted their first step single stance duration by step type. Generally, first step single stance duration did not differ between fallers and non-fallers for any step type except for MS at-BTL.

3.6. First step length

First step length was not different between groups ($F_{1,27.6} = 0.28$, $p = 0.6035$; fallers: 19.8 (1.2), non-fallers: 20.6 (0.8) % of height), but became longer with larger perturbation magnitude ($F_{1,103} = 52.57$, $p < 0.0001$; above-BTL: 22.8 (0.8) %, at-BTL: 17.6 (0.8) %), and varied by step type ($F_{3,113} = 184.98$, $p < 0.0001$; LSS: 21.4 (1.1), CFS: 28.1 (1.1), CBS: 24.3 (0.9), and MS: 7.0 (0.9) %) with MS having the shortest step length. Group by step type interactions were significant above-BTL ($F_{7,121} = 57.64$, $p < 0.0001$) and at-BTL ($F_{7,122} = 37.59$, $p < 0.0001$). Fig. 5 shows the results from post-hoc analyses that both fallers and non-fallers varied their first step length by step type (see bracket comparisons for each group). There were no group differences observed for any step type.

3.7. Trunk angular position relative to the gravity line at first step landing

Overall, trunk angular position relative to the gravity line at first step landing was not different between groups ($F_{1,24.2} = 2.72$, $p = 0.1118$; fallers: 10.5 (1.0), non-fallers: 8.7 (0.5) degrees), or by perturbation magnitude ($F_{1,76} = 2.77$, $p = 0.1001$; above-BTL: 9.3 (0.6), at-BTL: 10.0 (0.6) degrees), but differed by step type ($F_{3,81.1} = 4.80$, $p = 0.0040$; LSS: 9.7 (0.7), CFS: 10.0 (0.7), CBS: 8.4 (0.6), and MS: 10.3 (0.6) degrees) with MS having the largest trunk position. Group by step type interactions were significant above-BTL ($F_{7,114} = 4.31$, $p = 0.0003$) and at-BTL ($F_{7,115} = 3.14$, $p = 0.0045$). Fig. 6 shows the results from post-hoc analyses that, only non-fallers who altered their trunk position by step type (see bracket comparisons) while fallers did not. Group differences above-BTL were significant for CBS and MS, indicating that group differences were better demonstrated above-BTL.

4. Discussion

This is the first comprehensive investigation comparing the different types of medio-lateral protective stepping response strategies following lateral perturbations of standing balance in older adult fallers and non-fallers. Although BTL was not different between groups, differences in stepping performance indicated that: 1) Fallers stepped more often used MS while non-fallers mainly used CBS; 2) non-fallers had an overall lower step count and only non-fallers significantly modified the step count by step type; 3) no group differences were identified for first step spatio-temporal kinematic parameters, when averaged over the four step types. Group differences were only identified when examined between different step types. Moreover, non-fallers better modulated their stepping parameters by step type at the critical instants of first step onset (timing) and landing (trunk angular position). In contrast, ongoing step execution parameters (single stance duration and step length) were comparable between fallers and non-fallers with both adaptively modifying their stepping metrics by step type; and 4) overall, group differences were generally better demonstrated above-BTL.

4.1. BTL

We had expected that fallers would demonstrate a lower BTL than non-fallers, however, this was not observed. It is possible that since the perturbation threshold that evokes stepping in response to the waist-pulls is a function of both the perturbation displacement and velocity (Mille et al., 2003), the limited sets of magnitudes applied in the current study may have been too coarsely incremented to sensitively differentiate between the groups.

4.2. Step count and step type use patterns differentiate fallers and non-fallers

The finding that fallers had an overall higher step count is consistent with previous studies using lateral perturbations and is among the most potent stepping predictor variables for estimating the future risk of falls (Hilliard et al., 2008; Mille et al., 2013). This multiple-stepping behavior likely reflected, at least in part, a less biomechanically stable first step response that required additional steps in order to secure balance recovery (Hilliard et al., 2008; Mille et al., 2013; Carty et al., 2012). Moreover, we also found that group differences in step count for most step types were better identified above-BTL. To our knowledge, this study is the first to show that fallers did not adaptively modify their step count for different types of stepping, suggesting that they might have had limited ability to modify their

stepping performance. Alternatively, they might have selected to consistently use multiple steps to better secure their balance recovery.

For both groups, we observed a much lower incidence of loaded-leg-stepping (i.e., LSS, <10% of all stepping responses) than unloaded-leg-stepping (i.e., CFS, CBS and MS combined), a finding that resembled previous reports (Mille et al., 2013; Yungheer et al., 2012). This result supported the contention that older adults predominantly use unloaded-leg-stepping, presumably to take advantage of the mechanical effects of lateral perturbations that passively moves the CoM relative to BoS, thereby assisting with the active weight transfer prior to stepping (Mille et al., 2005, 2013; Yungheer et al., 2012). Despite the relatively low occurrence of loaded-leg-stepping trials for both groups, LSS can differentiate fallers from non-fallers whereby fallers show a greater mean step count for such trials. LSS is considered the most effective stepping strategy biomechanically among the four step types analyzed because LSS is always associated with widened BoS and higher torque generation (Patton et al., 2006). This may indicate that fallers have limited ability to execute biomechanically more effective LSS that possibly related to a reduced hip abductor–adductor power production capacity (Inacio et al., 2014) required for taking LSS.

4.3. Why do fallers frequently use medial-steps?

Regarding the use of different sub-types of unloaded-leg-stepping, previous studies comparing younger and older adults have reported either no difference in the frequency between crossover-steps and side-step-sequence (Maki et al., 2000) or a similar frequency of crossover-steps (young: 61%, old: 68%) in response to lateral waist-pulls (Sturnieks et al., 2012). Our results showed that older non-fallers had a similar frequency of using crossover-steps (~60%, CFS and CBS combined) as reported previously (Sturnieks et al., 2012), but older fallers had a greater frequency of using MS (especially above-BTL, >50%). We speculate that reduced somato-sensation often associated with increased fall risk with older age (Allet et al., 2014), may partially explain fallers' more frequent use of MS as it has been shown that hypothermic anesthesia of the foot soles changed the most frequent stepping type from crossover-steps to medial-steps in response to lateral platform perturbations even among younger adults (Perry et al., 2000). By more often taking MS, older fallers took advantage of passive limb unloading involving a quicker step onset (Maki et al., 1996, 2000; Mille et al., 2005, 2013; Yungheer et al., 2012) and avoided the longer and more complex step trajectory associated with crossover-stepping (Maki et al., 1996, 2000; Perry et al., 2000). That fallers more often used MS over the other step types may also suggest that MS is less demanding to control (Maki et al., 2000) and/or may be an early abortion of crossover-stepping (Perry et al., 2000). Although there may be some advantages for fallers to use MS, MS is considered biomechanically the least effective stepping strategy of the four step types analyzed because MS is always associated with a shorter initial step length that limits the BoS adjustment and effective torque generation (Hsiao-Wecksler and Robinovitch, 2007), and involves greater trunk motion that may further destabilize balance (Mille et al., 2005; Hurt et al., 2011). Therefore, any advantages of taking MS may be offset by its limitations in stabilizing balance recovery.

4.4. Fallers did not modulate first step characteristics at step onset and landing for different step types

In contrast with non-fallers, fallers did not adaptively modify their first step spatio-temporal parameters for different step types when initiating and terminating the first step. These instants are particularly vulnerable points in the stepping continuum due to the abrupt transitions that occur in the BoS configuration between bi-pedal and uni-pedal stance relative to the moving body CoM. Although the lack of first step modulation for different step types has not been previously reported, a similar lack of adaptation of stepping responses between forward and lateral perturbations has been observed in fallers compared with non-fallers (Mille et al., 2013).

There are several possible explanations for the lack of step onset time modulation for different step types in the fallers. First, fallers may have made the decision to step in advance based on the occurrence of the waist-pulls rather than on waiting to approach their mechanical limits of stability (Pai et al., 1998). Thus, if the step onset time in fallers is more determined by a conscious decision to step in response to the perturbation, then step onset timing will be less likely to vary between step types regardless of differences in the evolving state of mechanical instability. Second, fallers may have relied on the passive mechanical unloading provided by lateral perturbation, thus leading to a non-differentiated step onset time for different step types. Third, a greater impairment in leg somato-sensation may explain why fallers may have over-relied on the passive unloading effects to take steps.

It is noteworthy that fallers can perform different step types even though they do not modulate the initiation timing for the different step types. We reasoned that a relatively longer duration of step execution may allow fallers more time to better utilize on-line processing of all available sensory information reflecting the evolving state of instability for the control of stepping. However, on first step landing, faller's diminished somato-sensation could conceivably have again posed challenges in producing quick and appropriate responses leading to un-differentiated trunk angular position control for different step types.

4.5. Effects of perturbation magnitude on differentiating fallers and non-fallers

Our results also showed that group differences are better demonstrated for perturbation magnitudes above-BTL for several step types with respect to stepping frequency, step count, and trunk position at first step landing. Therefore, it is recommended that using perturbation magnitudes that are likely to induce multiple steps in future studies will more sensitively differentiate balance performance abilities between fallers and non-fallers.

4.6. Clinical implications

Although our results demonstrate that two variables, step onset time and trunk angular position, at critical time points can differentiate between fallers and non-fallers, reliable group discrimination is only demonstrated by examining their adaptive modulation between different step types and not when averaged-across or within each step type. However, these two variables have potential to serve as useful outcome measures for fall intervention programs because individuals appear to be able to adaptively modify recovery responses even after a short practice session (i.e., after practice, step onset time became longer with

emergence of anticipatory postural response (McIlroy and Maki, 1995) indicating active control rather than taking advantage of passive limb unloading involving a quicker step onset. Trunk excursion also decreased quickly after practice (Hurt et al., 2011)). The lack of consistent group discrimination by step type in our results might be due to the limited number of available stepping trials for each step type, a statistical power issue that can be addressed by including a larger number of stepping trails in intervention studies.

We have previously proposed that improved predictive motor control (Yungher et al., 2012) may partially explain the observed short-term adaptations. This view is consistent with findings that sensory inputs can modify step initiation (Perry et al., 2000). However, adaptations contributed by improvement in neuromuscular performance cannot be ruled out especially for longer-term intervention studies.

5. Limitations

The findings from this study conducted with a cohort of community-dwelling older people who were relatively healthy and functionally independent may not be directly applicable to frail older people with greater balance impairment. Nevertheless, understanding the complexities associated with medio-lateral stepping remains useful in elucidating balance control mechanisms that can help to guide falls prevention programs and intervention development. Similarly, results of protective stepping in response to waist-pulls may not be generalized to other types of fall provoking perturbations. However, regulating the relationship between the body's CoM and BoS such as by stepping, is a fundamental means of maintaining balance in both real-life and laboratory situations involving externally applied perturbations. Therefore, it is likely that differences between older fallers and non-fallers in their abilities to compensate for balance perturbations, reside in their control of protective stepping responses rather than in the means of perturbations that induce falling.

6. Summary

In summary, fallers primarily used biomechanically less effective medial-stepping in response to lateral perturbations of balance with associated higher step counts. They also demonstrated a lack of modulation of stepping parameters at the critical instants of first step initiation and termination where transitions in the BoS configuration occur. Further studies investigating the contributions of specific sensorimotor and muscle performance deficits with older age and other clinical conditions that impair protective stepping can elucidate underlying mechanisms and provide insight for falls prevention interventions.

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Fig. 1.

Four types of first step protective stepping responses. Rear-view of stick-figures plotted from kinematic data of an older faller. → indicates waist-pull to the right side. The dotted line is the stepping leg. — indicates the leg position behind the other leg.

Fig. 2.

Mean step count (standard error). Post-hoc analyses for significant group by step type interactions above- and at-the Balance Tolerance Limit (BTL). Only non-fallers significantly varied their step count for different step types. Significant pair-wise comparisons between step types are indicated for non-fallers by bracketed comparisons. Significant group differences for each step type are shown by symbols between two adjacent bars (e.g., ** indicates group comparison for LSS above-BTL). Significant group differences were observed for LSS and CFS above-BTL, and for LSS at-BTL. Symbols for p (or adjusted) values: ** < 0.01, * < 0.05, and ? < 0.10.

Fig. 3.

Mean first step onset time (standard error). Post-hoc analyses for significant group by step type interactions above- and at-BTL. Only non-fallers altered their step onset time by step type, with significant pair-wise comparisons between step types indicated by bracketed comparisons. A significant group difference was observed for MS at-BTL. Other notations are the same as in Fig. 2.

Fig. 4.

Mean single stance duration (standard error). Post-hoc analyses for significant group by step type interactions above- and at-BTL. Both fallers and non-fallers altered their single stance duration by step type, with significant pair-wise comparisons between step types marked for fallers (gray lines) and non-fallers (black lines). Symbols for p (or adjusted values): **** 0.0001; *** 0.001. Other text and symbol notations are the same as in Fig. 2. A significant group difference was observed for MS at-BTL.

Fig. 5.

Mean first step length (standard error). Post-hoc analyses for significant group by step type interactions above- and at-BTL. Both fallers and non-fallers altered their step length by step type, with significant pair-wise comparisons between step types indicated for fallers and non-fallers by bracketed comparisons for each group. Text and symbol notations are the same as in Fig. 4.

Fig. 6.

Mean trunk angular position (standard error) relative to the line of gravity at first step landing. Post-hoc analyses for significant group by step type interactions above- and at-BTL. Only non-fallers altered their trunk position by step type, with significant pair-wise comparisons between step types indicated by bracketed comparisons. A significant group difference was observed for CBS and MS above-BTL. Text and symbol notations are the same as in Fig. 4.

Table 1

Numbers of participants with BTL at each waist-pull magnitude.

	Fallers	Non-fallers
BTL at magnitude-5	0	0
BTL at magnitude-4	2	6
BTL at magnitude-3	7	22
BTL at magnitude-2	7	7
BTL at magnitude-1	0	1
Total participant number	16	36

Table 2

Percentage of trials for each step type, and for trials with no stepping.

	LSS	CFS	CBS	MS	No stepping	Total number of trials
Above-BTL						391
Fallers	5.9% ****	14.6% ****	29.4% ****	50.1% ****	0%	
Non-fallers	8.9% ****	17.5% ****	43.6% ****	30.0% *	0%	829
At-BTL						189
Fallers	8.4% ****	18.5% *	28.6%	37.6%	6.9%	
Non-fallers	7.4% ****	17.9% ****	43.4% ****	27.9% ****	3.4%	408
Below-BTL						256
Fallers	6.3%	0%	7.4%	7.8%	78.5%	
Non-fallers	3.1%	0.5%	4.2%	10.1%	82.1%	742

LSS: lateral-side-step; CFS: crossover-front-step; CBS: crossover-back-step; MS: medial-step. Shaded cells indicate the most frequently used step type for each group at specified perturbation magnitude (i.e., each row) (below-BTL not analyzed). Superscript symbols indicate significant difference in stepping frequency between each step type and the most prevalent step type (i.e., for fallers, each step type compared to MS; for non-fallers, each step types compared to CBS)

= $p < 0.0001$;

= $p < 0.001$;

*
= $p < 0.05$; p adjusted.

Bold-text cells indicate significant difference in stepping frequency between fallers and non-fallers for that step type at specified perturbation magnitude.