Journal of Biomechanics 48 (2015) 963–968

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Center of pressure control for balance maintenance during lateral waist-pull perturbations in older adults

Masahiro Fujimoto, Woei-Nan Bair, Mark W. Rogers *

Department of Physical Therapy and Rehabilitation Science, University of Maryland School of Medicine, Baltimore, MD 21201, USA

ARTICLE INFO

Article history:
Accepted 4 February 2015

Keywords:
Falls
Balance
Perturbation
Center of pressure
Center of mass

ABSTRACT

When balance is disturbed, location of the center of pressure (COP) contributes to a person's ability to recover from a perturbation. This study investigated COP control prior to first step lift-off (FSLO) during lateral perturbations in older non-fallers and fallers. 38 non-fallers and 16 fallers received lateral waist-pulls at 5 different intensities. Crossover stepping responses at the intensity level where the largest number of subjects responded with crossover steps were analyzed. Whole-body center of mass (COM) and COP positions in the medio-lateral (ML) direction with respect to the base of support (BOS), and COP velocity were calculated. An inverted pendulum model was used to define the BOS stability boundary at FSLO, which was also adjusted using the COP position at FSLO (functional boundary). No significant differences were found in the COP velocities between fallers and non-fallers ($p > .093$). However, the COP positions for fallers were located significantly more medial at FSLO ($p \le .01$), resulting in a significantly reduced functional boundary. Although the stability margins, measures of stability based on the BOS, were significantly larger than zero for fallers ($p \le .004$), they were not significantly different from zero for the functional boundary, i.e., reaching the functional stability limit. Fallers had reduced functional limits of stability in the ML direction, which would predispose them to more precarious stability conditions than non-fallers. This could be a cause for taking more steps than non-fallers for balance recovery as we observed. The functional boundary estimation may be a more sensitive marker of balance instability than the BOS boundary.

1. Introduction

Falls are the leading cause of serious injuries in older people due to age-related declines in balance control (Nevitt et al., 1989; Tinetti et al., 1988). An impaired ability to control lateral balance is an important aspect of balance problems contributing to falls (Maki and McIlroy, 2006; Rogers and Mille, 2003). A directional vulnerability to falling sideways among older individuals has been supported by a previous prospective study (Hilliard et al., 2008), recent experimental findings on multi-directional protective stepping (Mille et al., 2013), and an observational surveillance study of real-life falls (Robinovitch et al., 2013).

In order to effectively recover balance when standing stability is perturbed, protective steps must be appropriately timed and adjusted to arrest the motion of the whole-body center of mass (COM) (Mille et al., 2013). Older adults are much more likely than younger adults to take multiple balance recovery steps (Luchies et al., 1994; Maki et al., 2000; Mille et al., 2013, 2005), where an inability to recover lateral balance with a single step is predictive of future falls (Hilliard et al., 2008). Moreover, younger adults more often use a side step maneuver with the limb that is passively loaded by the lateral perturbation, whereas older adults more frequently use crossover steps with the passively unloaded limb (Mille et al., 2013, 2005). While crossover stepping with the passively unloaded leg facilitates the onset of stepping, it increases the potential for inter-limb collisions and subsequent falls (Maki et al., 2000; Mille et al., 2005). Thus, balance recovery steps for older adults appear to be less efficient using multiple steps and less effective with more inter-limb collisions using crossover strategies.

While protective stepping parameters after first step lift-off (FSLO), such as step count and step type, could be used as measures of dynamic balance function, they do not fully capture the evolving state of balance stability represented by the COM-base of support (BOS) relationship. For example, the location of the center of pressure (COP) prior to FSLO contributes to a person's ability to recover balance from a perturbation. When standing balance is disturbed, acceleration (or deceleration) of the COM is directly related to the distance between the COP and COM (Winter et al., 1998). Since the BOS provides a possible area for COP movement, the BOS boundaries have been considered as stability limits within which balance is maintained by

* Correspondence to: Department of Physical Therapy and Rehabilitation Science, University of Maryland School of Medicine, 100 Penn Street, Allied Health Building 205D, Baltimore, MD 21201, USA.
Tel.: +1 410 706 0841; fax: +1 410 706 6387.
E-mail address: MRogers@som.umaryland.edu (M.W. Rogers).

http://dx.doi.org/10.1016/j.jbiomech.2015.02.012
0021-9290/

rapidly moving the COP to keep the COM from going outside the BOS (Hof et al., 2005; Pai et al., 1998; Winter, 1995). Therefore, how "fast" and how "far" the COP moves with respect to the BOS prior to FSLO is importantly involved with dynamic balance control.

In addition to the COM–BOS relationship, balance would not be maintained if the COM has a sufficiently large horizontal velocity (Brown et al., 1999; McIlroy and Maki, 1996; Pai and Patton, 1997; Pai et al., 1998). Thus, dynamic balance stability has been quantified based on the position–velocity relationship between the COM and BOS (Carty et al., 2011; Hof et al., 2005; Pai and Patton, 1997; Pai et al., 1998), where an analysis applied to the frontal plane showed that older adults used a conservative strategy by stepping well before their stability limit was reached (Patton et al., 2006).

While these models used the BOS boundaries as limits of COP control, it has been shown that the functional limit of the BOS (FBOS), defined as the effectively utilized area for COP movement, is decreased with aging (Fujimoto et al., 2013; King et al., 1994). Such a reduced FBOS would limit an individual's ability to maintain balance because the COP–COM distance is proportional to the COM acceleration (Winter et al., 1998), which is important for regulating the momentum induced by perturbations as described earlier. A reduced area used for COP movement could predispose older individuals to a precarious condition for maintaining balance stability.

To further address these issues, the objective of this study was to investigate the COP control prior to and at FSLO during crossover protective stepping in response to lateral perturbations of standing balance in older non-fallers and fallers. COP velocity prior to FSLO and COP position at FSLO were calculated to assess the COP control. An inverted pendulum model was used to define the BOS lateral stability boundary at FSLO, which was also adjusted using the COP position at FSLO (functional boundary). We hypothesized that fallers would demonstrate a reduced functional boundary with a slower COP velocity than non-fallers.

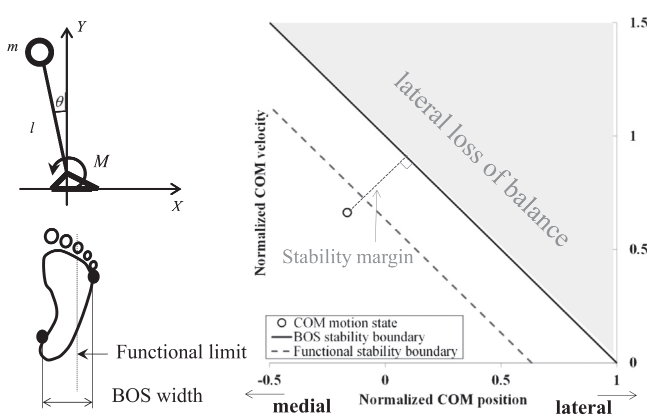

Fig. 1. A single-link-plus-foot inverted pendulum model in the frontal plane and lateral stability boundaries. X indicates the COM position in the medio-lateral direction. *m*, *l* and *M* are whole body mass, pendulum length (distance from the ankle to the COM), and ankle joint moment. The lateral stability boundaries were defined in two ways: one with the BOS width (BOS stability boundary), and the other adjusted with the COP position at FSLO, considering it as a functional limit for COP movement (functional stability boundary). Stability margins were calculated as the shortest distances from the experimental data to those stability boundaries.

2. Methods

2.1. Subjects

Thirty-eight healthy, community dwelling older adults [non-fallers: 19 men/19 women; age: 74.1 (SD 7.5) years; height: 1.67 (SD 0.09) m; body mass: 76.2 (SD 14.2) kg], and 16 healthy older adults with a history of falls [fallers: 6 men/10 women; age: 72.9 (SD 4.6) years; height: 1.67 (SD 0.11) m; body mass: 80.0 (SD 19.2) kg] participated in this study. Any individual who fell one or more times in the year prior to testing was categorized as a faller (Lord et al., 1999). A fall was defined as "coming to rest unintentionally on the ground or lower level, not as a result of a major intrinsic event (such as stroke) or overwhelming hazard" (Tinetti et al., 1988). Participants were medically examined by a physician to assess exclusion criteria including: (1) cognitive impairment (Folstein Mini Mental Score < 24); (2) sedative use; (3) non-ambulatory; (4) any clinically significant functional impairment related to musculoskeletal, neurological, cardiopulmonary, metabolic or other general medical problems; (5) participation in any regular vigorous or muscle strengthening exercise regimen; and (6) Centers for Epidemiological Studies Depression Survey score > 16. 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 Veteran's Administration Medical Center.

2.2. Data collection

Participants received a total of 60 randomly applied, position-controlled, motor-driven waist-pull lateral perturbations at five different intensities (Levels 1–5) in the left and right directions (L and R pulls). The system has been previously described (Pidcoe and Rogers, 1998) and used in prior studies (Hilliard et al., 2008; Mille et al., 2013, 2005; Young et al., 2013; Yungher et al., 2012). Participants wore a waist belt to which cables were attached and through which the perturbations were applied. Six trials were conducted for each intensity and direction (2 directions $\times$ 5 intensities $\times$ 6 repetitions). The order in which the trials were presented was randomized to minimize anticipation and sequence learning effects. Participants stood in a self-selected, comfortable standing position at the start of each trial with each foot on a separate force platform (AMTI, Newton, MA, USA). The foot locations were traced onto the platform surface to ensure consistent initial foot placement over the trials. Participants were instructed to "relax and react naturally to prevent themselves from falling."

Whole body motion was captured with a six-camera motion analysis system (Vicon 460, Oxford, UK). 28 reflective markers were placed according to Eames et al. with additional markers on the medial malleoli and 5th metatarsophalangeal (MP) joints (Eames et al., 1999). Three-dimensional marker trajectories were collected at 120 Hz and smoothed using a fourth-order Butterworth filter with a cut-off frequency of 8 Hz. Ground reaction forces (GRFs) were collected by two force platforms located under each foot at 600 Hz and filtered with a 10 Hz cut-off frequency (Hernandez et al., 2012; Maki et al., 1994).

2.3. Data analysis

Since crossover stepping is a common maneuver used by older adults to recover lateral balance (Maki and McIlroy, 2006; Mille et al., 2005), responses to the lateral waist-pulls at intensity Level 4, where the largest number of subjects responded with crossover steps (74% non-fallers (28/38) and 44% fallers (7/16) for L pulls, and 71% non-fallers (27/38) and 69% fallers (11/16) for R pulls), were

Table 1
Subjects' age, height, weight and stance width. Data are presented as mean (SD). Stance width was calculated as the mean distance between the left and right lateral ankles and 5th MP joints in the ML direction prior to the onset of perturbation. No significant differences were found between non-fallers and fallers ($p \geq .087$).

COP velocities	L pull			R pull		
# of subjects	Non-fallers 15 Men/13 women	Fallers 1 Man/6 Women	*p*	Non-fallers 16 Men/11 Women	Fallers 4 Men/7 Women	*p*
Age [years]	72.8 (7.1)	71.7 (4.9)	.635	74.1 (7.3)	72.6 (4.6)	.450
Height [m]	1.67 (0.08)	1.62 (0.04)	.087	1.68 (0.08)	1.69 (0.12)	.807
Weight [kg]	76.6 (14.2)	73.9 (15.8)	.653	77.6 (14.2)	82.7 (20.9)	.388
Stance width [cm]	33.6 (4.5)	35.0 (3.4)	.460	34.2 (4.4)	36.4 (3.5)	.160