Can initial and additional compensatory steps be predicted in young, older, and balance-impaired older females in response to anterior and posterior waist pulls while standing?

Brian W. Schulza,b,c,*, James A. Ashton-Millera,c, Neil B. Alexanderb,c,d

Abstract

The initiation of a single compensatory step in response to balance perturbations has been predicted with accuracies of up to 71%. We sought to determine whether similar methods also could be used to predict the onset of additional compensatory steps in both healthy and balance-impaired older females. Anterior and posterior waist pulls of five different magnitudes were applied to 13 unimpaired young (mean age 23 years), 12 unimpaired older (mean age 71 years), and 15 balance-impaired older (mean age 76 years) women. Body segment kinematic data were recorded at 100 Hz. A step was predicted when the time for the center-of-mass to reach the vertical projection of the boundary of the base-of-support fell below a certain threshold. The results show that 83% of all steps and non-steps were correctly predicted at an optimal time-to-boundary threshold ($t_{opt}$) of 0.78 s. Step prediction accuracy did not differ significantly by group: 86% of steps and non-steps by young, 84% by unimpaired old, and 82% by balance-impaired old women were correctly predicted at $t_{opt}$ of 0.58, 0.67, and 0.78 s, respectively. Anterior steps and non-steps were predicted more accurately than posterior ones (94% vs. 79% correct at $t_{opt}$ of 0.52 and 0.84 s, respectively) and initial steps were better predicted than additional ones (87% vs. 81% correct at $t_{opt}$ of 0.77 and 0.34 s, respectively). We conclude that this step prediction method reasonably predicts initial and additional steps in the anterior and posterior direction by all three subject cohorts.

Keywords: Stepping; Perturbation; Falls; Postural control; Aging

1. Introduction

A compensatory step is the most common method for preventing a fall in response to a perturbation of upright stance (Luchies et al., 1994; McIlroy and Maki, 1993; Rogers et al., 1996). The question of what mechanical stimuli the central nervous system uses to decide when to step remains controversial. Velocity-based dynamic models have proven to be more accurate than displacement-based static models for predicting a single step following a perturbation (Pai et al., 1998; Pai et al., 2000; Maki and McIlroy, 1999). However, a dynamic model has not been used to predict the initiation of steps beyond this initial step. The primary goal of the present paper is to evaluate an empirical method for predicting the occurrence of initial and additional steps.

Older individuals have been shown to initiate steps more readily and utilize multiple steps more often than younger and more able subjects (for example, Brauer et al., 2002; Luchies et al., 1994; McIlroy and Maki, 1996; Thelen et al., 1997; Mille et al., 2003). This may be due
to a reduced threshold for step initiation in older and more impaired subjects. Earlier work on compensatory stepping responses to waist-pull perturbations indicates that age and balance impairment-related differences in the initiation of initial and successive steps may be associated with the control of whole-body center-of-mass (COM) velocity (Schulz et al., 2005) and that this may differ in the anterior and posterior directions. Therefore, our secondary goals were to evaluate our dynamic prediction method in both the anterior and posterior directions, as well as the effects of age and balance impairment.

Several center-of-pressure (COP) variables, such as the sway amplitude, range, path length, area, and time–frequency analysis have been used to evaluate the displacement limits of static balance (Loughlin et al., 2003). The dynamic limits of static balance have also been evaluated by comparing the time required for the COP to reach the boundary of the base-of-support (BOS) given its current velocity and position. This prediction has been defined as the “time-to-boundary” of the COP with the stability boundary and it has been used to examine the dynamic aspects of postural sway in bipedal, non-stepping balance tasks (Slobounov et al., 1998; van Wegen et al., 2001; van Wegen et al., 2002). This evaluation of the dynamic aspects of static balance is noteworthy, but the construct validity of this method has not been tested. The COP moves within the BOS defined by the feet in order to “guide” the COM and thus maintain balance. This means that the approach of the COP to the BOS boundary does not imply biomechanical instability, but rather ankle torques being applied in the attempt to restore stability. Thus, this time-to-boundary concept is here applied to the COM in order to predict the initiation of compensatory steps. The COM time-to-boundary value (τ) reflects the present state of the body in relation to its environment, while we believe that the COP time-to-boundary reflects the control signal used to alter this state.

This study describes the application of the COM time-to-boundary concept to both initial and subsequent compensatory steps in young, older, and balance-impaired older females exposed to waist-pull balance perturbations. We hypothesized that the greater incidence of initial and additional step initiation in the older and balance-impaired older subjects was not due to more conservative thresholds of step initiation, but rather to differing abilities to resist perturbations.

2. Methods

The 13 unimpaired young and 12 older women were capable of unipedal stance times greater than 30 s on at least one of three trials, while all three unipedal stance time trials of the 15 balance-impaired older women were less than 10 s. All young completed a medical questionnaire while all older subjects were screened by a nurse practitioner prior to testing. While the unimpaired old had minimal findings on directed history and physical, one-third of the balance-impaired older women noted pain or limitations in motion of the hip or knee, nearly half noted problems with their balance, and, on exam, four (26%) had mild hip flexion weakness and two (13%) had a positive Romberg (failure to stand bipedally with eyes closed for at least 10 s).

Subjects wore their own shoes and stood in a comfortable position with each foot on a ground-level force plate. This position was replicated for all trials by tracing the initial outline of the feet on the ground. Care was taken to ensure that this position was returned to before each trial and that one foot was not in front of the other. Subjects wore a full-body safety harness that did not restrict postural responses and only provided support in the event of a fall. Ten optoelectronic markers were used on the left and right fifth metatarsal-phalangeal joints, lateral malleoli, lateral femoral condyles, greater trochanters, and acromion processes. Switchplates (approx. 1 m²) on the ground anterior and posterior to the force plates (side-by-side AMTI OR6-6, Watertown, MA) helped determine initial step landing times.

Waist-pulls were applied in either an anterior or posterior direction. The pulling apparatus consisted of a rigid, padded belt worn over the safety harness and two pulling towers located directly anteriorly and posteriorly to the subject. A weight pan in each tower was connected to the belt at the midline by thin inextensible steel cables with a load cell in series to determine pull onset time.

The pan upon which the weights were placed weighed 2.7 kg each and slid vertically up and down in ball-bearing tracks. The pull force was redirected horizontally at a height of 102 cm (approximately waist level) by ball-bearing sheaves. The apparatus permitted usual postural sway because the pan dropped 2–4 cm before applying the pull force. To minimize anticipation of pull direction or magnitude, presentation of the anterior and posterior loads of 1, 2, 3, 4, and 5% of body weight was randomized. Based on pilot studies, the range in pull magnitudes safely elicited steps in the balance-impaired as well as the healthy subjects. The weights were released by solenoid mechanisms when the subjects were in the center of their sway pattern. The weighted pan fell up to 30 cm and data were collected for 3 s. The data collection was started approximately 0.5 s before weight release. The direction and magnitude of the pull were masked because the pans in both towers dropped simultaneously.

Subjects were instructed to recover their balance in a natural manner, but to keep their arms crossed on their
Two trials for each weight and direction were presented to each subject in a fixed order in two randomized blocks of ten trials (20 trials total). If the arms became uncrossed before step liftoff, the trial was repeated at a random point later that block. If the arms became uncrossed after the initial step liftoff, it was still included in the analysis, but both of these errors occurred infrequently and were small in magnitude and thus unlikely to alter the results.

All data were sampled at 100 Hz. Custom Matlab (Natick, MA) software routines were used to process all data from an OPTOTRAK 3020 (Northern Digital, Waterloo, Ontario, Canada) motion capture system. The raw marker and load cell data were digitally filtered using a second-order low-pass Butterworth filter with forwards and backwards passes and a cutoff frequency of 6 Hz. The force plate data were similarly filtered using a cutoff frequency of 30 Hz.

A step was counted when both foot markers traveled over 2 cm. Step liftoffs were indicated by a 3% body weight force plate threshold for initial steps and a 0.4 m/s foot marker resultant velocity threshold for subsequent ones. Step landings were determined by foot switchplate contact for the initial step and the foot marker velocity threshold for subsequent ones. The use of foot marker velocities to determine steps was necessitated by steps that did not fully leave the ground—i.e. they slid along the floor. Thresholds for foot marker velocity, ground reaction force, and switch plate voltage were chosen to minimize incorrect step liftoffs and landings by inspection of many (i.e. 50–100) sample trials. About half of these sample trials were of balance-impaired older women, as they stepped most often and exhibited more erratic responses. Segment masses and COM locations were calculated from the individual subject anthropometry using established values (de Leva, 1996). The antero-posterior location of the head–arms–torso COM was calculated by determining the distance along the antero-posterior head–arms–torso vector that the head–arms–torso COM must be located at in order for the whole-body COM to be located at the average location of the COP during a 3 s quiet stance trial, given the head–arms–torso mass and the masses and COM locations of the other body segments. The arm masses and COM locations were neglected in the determination of the superior–inferior location of the head–arms–torso COM, as the close proximity of the crossed arms to the head–torso COM and much smaller relative mass of the arms would have little effect on the head–arms–torso COM location. The location of the whole-body COM was calculated using the foot, shank, thigh, and head–arms–torso segmental centers of mass. This step prediction method was evaluated based upon whether a step did or did not occur for every possible “step prediction” opportunity. Prediction opportunities for steps consisted of the interval from the beginning of the trial or prior step landing until the step liftoff. Prediction opportunities for non-steps consisted of the interval from the beginning of the trial or the prior step landing until the end of the trial. Thus, if the data collection stopped while a step was in the air, then the following step could not be predicted. Therefore, a trial with a single step that lands before the trial ends would have two step prediction opportunities (a step and a non-step), while a trial with a single step that is still in the air at the end of the trial would have a single step prediction opportunity (the step only) and even a trial with no steps would have a single step prediction opportunity (a non-step).

The time-to-boundary ($\tau$) was calculated by dividing the instantaneous distance from COM to the BOS boundary in the direction of the pull along the COM velocity vector by the magnitude of the instantaneous COM velocity (Fig. 1) throughout the entire trial. The vertical components of all distance and velocity values were neglected—i.e. only horizontal plane positions and

![Fig. 1. General definition of time-to-boundary ($\tau$). See text for specific details.](image-url)
motions were considered in the calculation of $\tau$. If the COM velocity vector was ever in the opposite direction of the pull, the value of $\tau$ at that instant was assumed to be infinite.

A step was predicted to occur if $\tau$ fell below a specified threshold ($\tau_{\text{opt}}$) (see Fig. 2 for sample trials) and a non-step was predicted if $\tau$ never fell below this threshold. For example, a step would always be predicted to occur when $\tau \leq 0$ s while a step would never be predicted to occur when $\tau \to \infty$ (i.e. will never intercept BOS boundary). The $\tau$ threshold that resulted in the greatest percentage of correct predictions of steps and non-steps was determined to be the optimal time-to-boundary threshold, $\tau_{\text{opt}}$. Specific $\tau_{\text{opt}}$ were determined for each subject group, pull direction, and initial or additional steps. An ‘overall’ $\tau_{\text{opt}}$ (calculated across three groups, two pull direction, and all steps) was also determined.

The performance of this step prediction method was quantified in truth table form—i.e. a true positive was a step predicted when a step actually occurred; a true negative was a non-step predicted when a step did not occur; a false positive was a step predicted when a step did not occur; and a false negative was a non-step was predicted when a step actually occurred.

Group differences in age, height, weight, and body mass index were compared using one-way analysis of variance (ANOVA) and Bonferroni post-hoc multiple comparisons. Group differences in initial (pre-pull) foot placement were tested with a repeated-measures analysis of variance where group was the between-subject factor and trial order was the within-subject factor and the differences in antero-posterior and medio-lateral foot placement (as calculated from the mean of the foot marker position) at pull onset were the dependent variables.

Variations in model performance by group were also tested using an ANOVA, where percentage of correct predictions (i.e., true positives + true negatives) was the dependent variable. Variations in model performance by step direction and step number (initial or additional) were quantified using a paired $t$-test. For the group analysis, the data were collapsed across all trials and

Fig. 2. Time-to-boundary ($\tau$) profiles for two maximal posterior pull trials of unimpaired old women. No steps occurred for one trial (dashed curves) and two steps occurred for the other (solid curves). The top panel shows $\tau$ and the lower panel shows the numerator (left axis and thin lines) and denominator (right axis and thick lines) of $\tau$. The horizontal dashed line in the top panel represents the optimal time-to-boundary threshold ($\tau_{\text{opt}}$) of 0.93 s to base-of-support boundary for all posterior steps by unimpaired old. The arrows indicate the first time $\tau$ falls below this threshold (thus predicting a step) and the shaded areas in both panels indicate the actual step (between step liftoff and landing).
pull magnitudes to yield a single value for the percentage of correctly predicted steps or non-steps for each subject. The direction (anterior vs. posterior) and number (initial vs. additional) were similarly collapsed, although each subject had two values. The percentage of correctly predicted steps and non-steps was calculated and tested for the specific (by group, direction, and initial or additional step) and overall (for all groups, directions, and steps) optimal time-to-boundary values. SPSS and SAS (v10.1, SAS Institute Inc., Carey, NC) were used for all statistical analysis and \( p < 0.05 \) was considered statistically significant.

3. Results

Height and weight did not significantly differ between groups (Table 1), although body mass index did \( F(2, 37) = 4.24, p = 0.022 \), particularly between young and balance-impaired older women \( p = 0.021 \). The age difference between unimpaired and balance-impaired older women was not significant \( p = 0.093 \).

Of the ten pulls applied to each subject in each direction, the young and unimpaired old rarely took more than one step while the balance-impaired old frequently took more than one step (Table 2). The number of step prediction opportunities (Table 3) follows a similar pattern as the number of steps.

Despite the group differences in stepping (Table 2), there were no significant differences in step prediction accuracy or \( t_{opt} \) by group (Table 3 and Fig. 3, \( F(2, 37) = 0.42, p = 0.7 \) for group-specific and \( F(2, 37) = 0.18, p = 0.8 \) for overall \( t_{opt} \)).

Ninety-four percent of all anterior stepping behavior and 79% of all posterior stepping behavior were correctly predicted by the specific \( t_{opt} \) (Table 4 and Fig. 4). The overall shapes of the prediction accuracy-threshold curves were similar, but a higher percentage of correct anterior step predictions were evident \( t = 8.65, p < 0.0001 \) for directionally specific and \( t = 4.87, p < 0.0001 \) for overall \( t_{opt} \) along with lower \( t_{opt} \) (Table 4 and Fig. 4). The greater prediction accuracy in the anterior direction is shown by the greater separation between steps and non-steps as indicated by the mean COM velocity magnitude and COM distance to the BOS boundary at \( t_{opt} \) (Fig. 5).

Additional steps were less accurately predicted than initial steps (81 vs. 87%, \( t = 2.30, p = 0.027 \) for initial/additional-step-specific, and 75 vs. 87%, \( t = 2.77, p = 0.009 \) for overall optimal \( t_{opt} \); Fig. 6). \( t_{opt} \) was higher for initial steps than for additional

| Table 1 | Mean (SD) subject characteristics |
|-----------------|-----------------|-----------------|
| Number of women | 13               | 12              | 15              |
| Mean age (years)| 23 (3.6)         | 71 (5.6)        | 76 (6.3)        |
| Age range (years)| 18–29           | 64–80           | 65–84           |
| Height (m)       | 1.67 (0.08)      | 1.61 (0.08)     | 1.61 (0.06)     |
| Weight (kg)      | 61.0 (6.1)       | 61.8 (10)       | 66.7 (11)       |
| Body mass index (kg/m²) | 22.1 (3.2) | 23.4 (3.6) | 25.8 (3.4) |
| One-legged stance time (s) | >30 | >30 | 5.76 (3.3) |

aValue shown is the best effort of three attempts.

| Table 2 | Number of trials with \( N \) steps per trial, arranged by pull direction and group |
|-----------------|-----------------|-----------------|-----------------|-----------------|-----------------|
| \( N \)         | Anterior pulls  | Posterior pulls |
|                 | Young women     | Unimpaired old women | Balance-impaired old women | Young women     | Unimpaired old women | Balance-impaired old women |
| Initial 1       | 10              | 11              | 17              | 48              | 25              | 23              |
| Additional 2    | 0               | 2               | 17              | 5               | 13              | 35              |
| Additional 3    | 0               | 1               | 3               | 0               | 5               | 10              |
| Additional 4    | 0               | 0               | 2               | 0               | 1               | 3               |
| Additional 5    | 0               | 0               | 0               | 0               | 0               | 2               |
| Total additional steps | 0              | 3               | 22              | 5               | 19              | 50              |

steps (0.77 vs. 0.34, respectively; Table 4). The lower prediction accuracy for additional steps was primarily due to an increase in false positives for additional steps (Table 5).

4. Discussion

Steps were predicted based on a measure of the amount of time before subjects could no longer control
their balance with their current BOS, given their instantaneous position and velocity. The results show that 83% of all steps and non-steps were predicted at $t_{opt} = 0.78$ s. Using different methods, Hof et al. (2005) found temporal stability values of 0.4–14.1 s, so our $t_{opt}$ values of 0.34–0.80 s corroborate their findings. Thus, this step prediction method appears to be applicable to initial and additional steps in the anterior and posterior direction by all three subject cohorts, as long as $t_{opt}$ was based on a sufficient number of trials by subject group (i.e. > 100 stepping opportunities).

The accuracy of this step prediction method (83%) is similar to the accuracy of prior models (65% for (Pai et al., 1998), and 71% for (Pai et al., 2000)), but direct comparisons were not possible due to methodological differences. Unlike those two models, which used the kinematic data at the endpoint of the applied perturbation waveform to predict a single step, the present method required the calculation of a $\tau$ throughout the trial in order to predict initial and additional steps; also, it was tested using a perturbation that was continuously applied throughout the duration of the trial (i.e. it had no endpoint). While this method could have been directly compared to the static model by determining if the COM ever passed beyond the BOS boundary before step liftoff, a dynamic model has already been established to predict stepping better than the static one (Pai et al., 1998).

This step prediction method begins with the idea that the central nervous system requires a minimum amount of time to initiate a step, but also that a safety margin is employed (Pai et al., 1998) that causes steps to be initiated earlier than absolutely necessary. In other words, a conscious or automatic decision is made to step
at some point dictated by the intrinsic and extrinsic demands and capabilities of the situation and person. The identification of this point and how it is altered by age, balance impairment, perturbation direction, and multiple stepping was one of our goals.

Despite the large differences in age and unipedal balance times in these three groups, group differences in step prediction accuracy were not meaningful. This corroborates the findings of Pai et al. (1998) and contradicts the findings of Mille et al. (2003) who found significant age effects in step thresholds. However, their subjects were instructed to "try not to step" while the subjects in this study and in Pai et al. (1998) were instructed to respond naturally. We believe this change in instructions elicited maximal capacities rather than natural responses; so while the capacities of the young and old to resist perturbations may indeed differ, their natural responses to them do not.

The fact that plantarflexor strength exceeds dorsiflexor strength (Thelen et al., 1996), forefoot length exceeds rearfoot length, and the range of hip flexion exceeds hip extension all favor anterior pulls being more readily resisted than posterior pulls. This may explain the less conservative (i.e. lower) $t_{opt}$ and greater prediction accuracy in the anterior direction (Fig. 4 and Table 4). Regardless of the root cause, the directional difference in $t_{opt}$ indicates that posterior steps are initiated twice as early as anterior steps. The pulls applied for this study were generally not that difficult to resist and we believe that fear-based responses were not frequently elicited. Thus, we believe that the relative weakness of the dorsiflexors as compared to the plantarflexors, and the longer moment arm of the toes as compared to the heel, probably played a greater role.

The increase in false positives (Table 5) and reduction in accuracy (Fig. 5) for additional steps were probably due to steps occurring after data collection had stopped, but we had no reasonable way of knowing if an additional step would have occurred after the data collection ended. Applying the pull force over a shorter
period of time (say, 0.5 s) and recording until the subject stops stepping would help eliminate this uncertainty. Although this step prediction method was less accurate for additional than for initial steps, it is unclear whether this was attributable to fundamental differences in the initiation of initial and additional steps or to the aforementioned methodological limitation.

This step prediction method is limited in that it cannot predict steps a certain amount of time before liftoff without losing the ability to predict steps with dual-contact periods that are shorter than the prediction window. Additionally, the smaller the number of trials used to determine \( \tau_{\text{opt}} \), the more “stairstepped” the accuracy vs. threshold curve becomes and the less specific and reliable the \( \tau_{\text{opt}} \). It is for this reason that \( \tau_{\text{opt}} \) was not determined for initial and additional steps by each group in each direction. Furthermore, only antero-posterior stability boundaries of the BOS (lines connecting the toes or heels) were considered here. Since most additional steps were in the posterior direction and lateral motion during posterior step landings (Schulz et al., 2005) would most likely intercept the BOS at the lateral margins of the foot, the inclusion of lateral stability boundaries may improve step prediction accuracy for additional steps, but at the cost of increased complexity. Additionally, although footwear was not standardized, we doubt that this introduced significant variability because all subjects wore athletic shoes. Finally, as all subjects were women, these results may not be reliably extrapolated to men.

We conclude that this step prediction method reasonably predicts initial and additional steps in the anterior and posterior directions by all three subject cohorts. It did this when \( \tau_{\text{opt}} \) was based on a sufficient number of trials (i.e. more than 100 stepping opportunities).

While young, unimpaired old, and balance-impaired old women took different numbers of initial and additional steps, they all chose to initiate steps at a similar biomechanical threshold for similar balance disturbances. This is not because the unimpaired and balance-impaired older women had different thresholds of step initiation when compared to the young women, but because the unimpaired and, especially, the balance-impaired older women could not resist crossing the same threshold as effectively.

In order to resist crossing the time-to-boundary threshold, future balance training interventions might consider training to lengthen steps and slow COM velocity more rapidly and effectively.

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References


