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Effects of step length on stepping responses used to arrest a forward fall

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Abstract

This study investigated effects of step length on the stepping response used to arrest an impending forward fall. Twelve healthy young (mean age 22, S.D. 3.3 years) males participated by recovering balance with a single step following a forward lean-and-release. Participants were instructed to step to one of three floor targets representing small, natural, and large step lengths. The effect of step length was examined on the primary outcome variables: pushoff time, liftoff and landing time, swing duration, balance recovery time, landing impulse, and center of mass (COM) characteristics. Pushoff and liftoff times were not affected by step length, although swing phase duration, landing and recovery times and the anterior-posterior (AP) impulse at landing increased with increasing step length. The results support the idea of an invariant step preparation phase. Given that our participants naturally chose not to utilize a step as short as they were capable of employing, healthy young individuals do not minimize recovery time nor strength requirements when selecting their step length.

Keywords: Balance; Motor control; Step length; Forward fall; Human

1. Introduction

Injurious falls are a significant health problem for older adults. Falls accounted for more than 14,000 deaths and 22 million medical visits in 1996 [1], and are estimated to account for 90% of hip fractures in older adults [2,3]. Even if an injury sustained after a fall is not serious, it can have a serious impact on an older adult's mobility, self-confidence, and independence [4]. Therefore, we need to understand the causes of falls in older adults to improve the methods used to identify, diagnose, and treat those at risk of falling.

A stepping response is often used following a balance perturbation to reconfigure the base of support (BOS) to encompass the body's center of mass (COM) [5]. Stepping responses used to arrest impending falls have been investigated in young and older adults with various perturbation techniques, including waist pulls, lean-and-releases, and platform movements. Luchies et al. [6], using backward waist pulls, found older adults stepped earlier, used a shorter initial step, and were more likely to employ a multiple step strategy instead of the single step strategy preferred by the young. McIlroy and Maki [7], using a platform perturbation, observed older adults tended to use a multiple step strategy. Thelen et al. [8], using a forward lean-and-release, demonstrated that age significantly reduced the maximum step length and largest lean angle from which a participant could regain balance using a single step. Rogers et al. [9], using a forward waist pull, found older adults stepped earlier and used a step with longer duration compared to young adults.

The multiple, small steps often used by older adults may be less effective in restoring balance compared to larger single step responses. For example, Hsiao and Robinovitch [10] observed that two-thirds of the elderly participants who fell after a backward lean-and-release attempted a multiple

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small step strategy. While there is evidence that small step lengths during gait initiation are associated with increased falling risk [11,12], the underlying causes for older adults to use shorter steps for balance recovery are not well understood. The biomechanical requirements for a smaller step, compared to a larger step, are decreased due to smaller body segment and joint rotations [6], lower peak torques at some joints [13], and reduced muscle activity [14]. Thus, older adults may choose a shorter step to reduce physical demands. Alternatively, older adults' shorter steps may be coupled with a tendency to initiate a step earlier in their response, and may represent a motor program initiated as fast as possible. Thus, this study examined the relationship between step length and step timing in young adults. We hypothesized that if young adults utilized a short single step response, they would initiate their response earlier similar to older adults. If, however, step initiation and step length were decoupled, then the young would modify step length without modulating step timing.

2. Methods

2.1. Participants

Twelve healthy young male participants (mean age 22, S.D. 3.3 years) participated after providing written informed consent approved by the institution's human subjects review board. All participants, recruited from university students and staff, denied significant head trauma, musculoskeletal impairments, and neurological disease. Foot dominance was determined by asking which foot he would use to kick a ball. Participants were paid for their participation.

2.2. Tasks

A sudden release from a static forward lean produced the fall-provoking disturbance [8,15]. We manipulated the step length by instructing the participant to recover their balance using a small, natural, and large sized step. Five trials were performed for each step length in a random order, resulting in 15 trials for each participant.

Prior to data collection, the participant assumed a comfortable standing posture on two adjacent force plates (one foot on each), barefoot, and arms crossed across the chest. A third force plate was placed anteriorly. Foot initial positions were traced onto clear contact paper covering each force plate and manually digitized. Practice trials were performed with the instructions to regain balance naturally using a single right foot step. The average toe landing position during the practice trials was determined and marked on the front force plate with tape. Tape was placed 10 cm in front of and behind the natural landing location to mark larger and smaller than natural landing locations, respectively. The tape locations were digitized and used to determine the error between the intended and actual step lengths. Before each trial, the desired right toe landing location was announced. Before data collection, practice trials (two for each landing location) were performed to familiarize the participant with test procedures.

The lean-and-release system consisted of a cable attached to a pelvic belt that supported the participant prior to release, a load cell to measure cable tension, a solenoid-activated hairline trigger designed to release the lean-control cable, and a microcontroller (Parallax Inc., Rocklin, CA, USA) to initiate data collection 500 ms before activating the cable release system. To insure that they were unaware of data collection initiation, participants listened to music through headphones. The cable length was adjusted until the cable supported 20% of the participant's body weight, which was large enough to insure all participants required a step response for balance recovery. The participants wore a safety harness, designed to prevent contact with the floor during a full fall, connected to an overhead frame through a load cell, which was monitored to indicate a failed recovery defined as a load exceeding 2.5% the participant's body weight.

2.3. Experimental measures

Motion and force data were synchronized and collected for 3 s with 100 Hz and 1000 Hz sampling rates, respectively. An Optotrak (Northern Digital Inc., Waterloo, Ont., Canada) measured right leg motions using infrared-emitting diodes (IREDs) attached to the right leg (second metatarsal, heel, lateral malleolus, lateral tibial epicondyle and tibial wand). Three force plates (Advanced Medical Technology Inc., Watertown, MA, USA) measured foot-support surface reactions for each foot independently at the initial location and the landing location of the right foot. A uniaxial load cell (Futek, Irvine, CA, USA) measured the safety harness load, and a biaxial custom-built load cell measured the lean-control cable tension. Force data were recorded on a personal computer using LabVIEW and a 16-bit A/D data acquisition card (National Instruments, Austin, TX, USA).

2.4. Data analysis

The step response was quantified using temporal parameters (pushoff time, liftoff time, landing time, and balance recovery time), kinematic parameters (step length, step error, and average speed), and kinetic parameters (landing force impulse and center of mass trajectories with respect to the base of support), which were derived from experimental measurements. Data from all trials were processed using MATLAB (Mathworks, Natick, MA, USA). Motion and force data were digitally low-pass filtered using a second order Butterworth filter with cutoff frequencies of 6 Hz and 30 Hz, respectively. Initial and final-time artifacts were minimized using forward and backward reflection of the data [16], and phase shift was eliminated by using forward and backward passes [17].

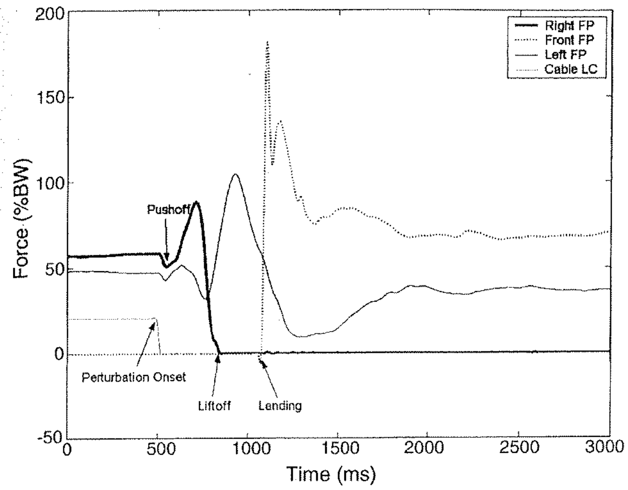

Fig. 1. Typical time history of right and left foot vertical forces, landing force, and lean-control cable load. Perturbation onset, pushoff time, liftoff time, and landing time are also depicted.

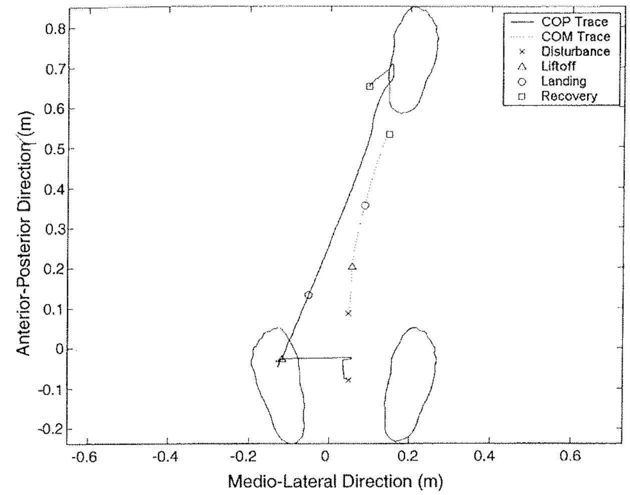

Fig. 2. Typical ground-plane projections of center of pressure (COP) and center of mass (COM) trajectories in relation to feet positions prior to perturbation and to right foot position at balance recovery.

The disturbance onset time and step foot pushoff, liftoff, and landing times were extracted from the force plate and load cell data (Fig. 1) using a threshold method. Disturbance onset was defined as the time when the cable tension dropped to zero. Pushoff time was defined as the time when the force exerted by the stepping foot initially decreased to a local minimum prior to increasing in preparation for weight transfer. Liftoff and landing times were defined as the times when the vertical force exerted by the stepping foot initially dropped below the threshold of 15 N, and then rose above 15 N, respectively (Fig. 1). Values of pushoff, liftoff, and landing times were expressed relative to the disturbance onset time. Swing phase duration was defined as the difference between liftoff and landing times. The temporal events were used to divide the response into three regions: double stance, single stance, and landing. Double stance is the region between disturbance onset and step foot liftoff time; single stance is the region between step foot liftoff and landing times (swing phase); and landing is the region between landing and balance recovery time.

Step length was defined as the distance between the position of the second metatarsal IRED at liftoff and landing times. The average step speed was calculated by dividing the step length by the swing phase duration. The toe's position upon landing was calculated by adding the distance between the toe and the second metatarsal IRED (measured prior to testing) to the anterior-posterior position of the IRED. Mean (S.D.) step lengths were 72.43 (5.12), 80.37 (5.11), and 90.28 (5.49) cm for the small, natural, and large step tasks, respectively. The large step length was significantly larger than the natural ($p < 0.001$) and small ($p < 0.001$) step tasks, and the natural step length was significantly larger than the small ($p < 0.005$) step task. Step error was the AP difference between the toe and target positions.

Center of pressure (COP) location was calculated using the foot-support surface reactions. Center of mass was calculated by double integrating the acceleration (calculated by dividing the force signal at each time step by the participant's mass) within each region [18], with the initial COM location calculated using a static equilibrium analysis about the ankle joints. Balance recovery time was defined as the time when the COM reached its maximum anterior position; i.e. the time at which forward momentum had been arrested. The AP distance between the COM position at recovery time and the edge of the base of support (based on the digitized foot outline, Fig. 2) was determined. The force impulse was calculated in all three directions by numerically integrating the force record with respect to time between landing and balance recovery times.

Of the 180 trials, trials were excluded from analysis if the participant stepped with the left foot (0 trials), did not step (0 trials), stepped on the right force plate after liftoff (0 trials), did not successfully land within 5 cm of the intended target (31 trials or 17.2%), or placed more than 2.5% body weight on the safety harness (12 trials or 6.7%).

2.5. Statistical analysis

Statistical analysis was accomplished using SPSS 9.0 (SPSS Inc., Chicago, IL, USA). A one-sample t-test was performed on the outcome variables within each task (small, natural, or large step) to determine if significant differences existed between left (four) and right (eight) leg dominant participants. Since no significant differences were observed, the two groups were pooled. Means for each outcome variable were calculated across the five trials for each task, and a one-way analysis of variance (ANOVA) was used to determine any significant between-task differences for the outcome variables. Tukey's post-hoc pairwise comparisons

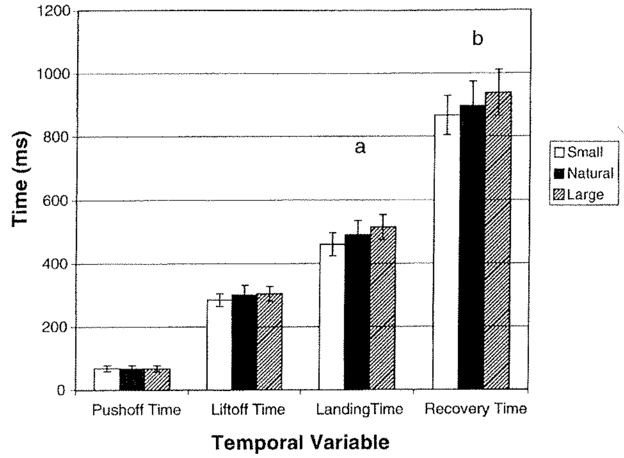

Fig. 3. Mean values of pushoff time, liftoff time, landing time, and balance recovery time for each task. Pushoff and liftoff times were not affected by task. (a) Landing time was smaller for the small step task compared to the large ($p = 0.006$) step task. (b) Balance recovery time was marginally smaller for the small step task compared to the large ($p = 0.05$) step task.

were performed to test for differences between pairs of tasks. A significance level of $p = 0.05$ was used for all analyses.

3. Results

3.1. Temporal parameters

Task had no effect on pushoff or liftoff time, but significantly affected swing time, landing time, and balance recovery time (Fig. 3). The large step task, compared to small, had significantly longer landing time ($p < 0.01$) and swing phase duration ($p < 0.005$). Recovery time was marginally shorter for the small step task compared to the large ($p = 0.05$) step task (Fig. 3).

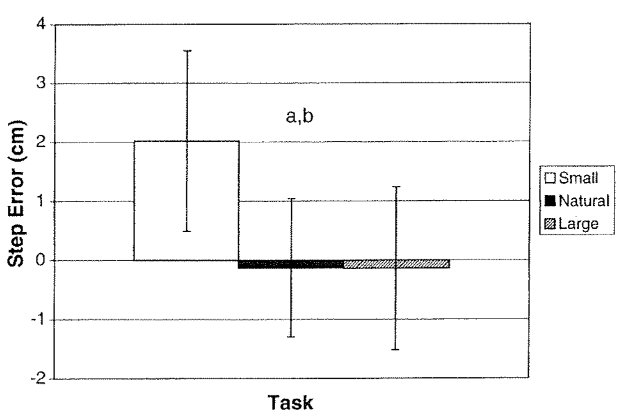

Fig. 4. Mean values of step error for each task. (a) Step error was significantly larger for the small step task compared with the natural ($p = 0.001$) step task. (b) Step error was significantly larger for the small step task compared with the large ($p = 0.001$) step task.

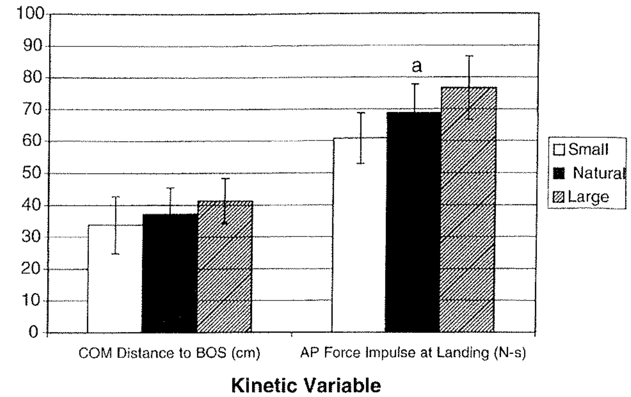

Fig. 5. Mean values of AP COM to BOS distance and AP force impulse for each task. COM-BOS separation was not significantly different among tasks. (a) Force impulse was significantly smaller for the small step task compared with the large ($p = 0.0001$) step task.

3.2. Kinematic parameters

Task had a significant effect on step error. The step error was significantly larger for the small step task compared to the natural ($p < 0.001$) and large ($p < 0.001$) step tasks (Fig. 4). Task had no effect on step speed.

3.3. Kinetic parameters

Task had no effect on the AP distance between the COM and the anterior limit of the BOS at recovery time. Task had a significant effect on AP force impulse at landing. AP impulse was significantly smaller for the small compared to the large ($p < 0.0001$) step task (Fig. 5).

4. Discussion

We tested our hypothesis that participants, when instructed to take a shorter step than natural, would initiate a step earlier in their response. Surprisingly, we found no significant adjustments of step pushoff or liftoff times when participants decreased or increased their step length by 10 cm compared to their natural step length. Instead, participants modulated the recovery portion of their response to successfully regain balance with different step lengths. This result suggests that the initiation of the balance recovery response is independent of the step length planned by the participant. This demonstrates a decoupling between step length and step liftoff time in young adults, and suggests that factors other than step length underlie the tendency of older adults to step earlier following a perturbation (e.g. Luchies et al. [6]). For example, older adults likely have a greater fear of falling [19], which could contribute to earlier steps often observed in older adults.

Our results support the idea of an invariant balance recovery preparation phase as proposed by Do et al. [15]. In that study, modulating lean angle resulted in participants adjusting step length rather than step liftoff time to recover balance. In the current study, modulating step length while maintaining the lean angle resulted in a similar invariant preparation phase (step pushoff or liftoff times).

Shortening the step length may have reduced the task's biomechanical demands. The small, compared to large steps, were characterized by a smaller AP force impulse during landing. Since AP momentum is directly related to the biomechanical strength necessary to produce a balance-restoring moment [20], it may be concluded that reduced strength requirements exist for smaller steps. Since the young were capable of using a short step but did not do so naturally, the young apparently do not minimize peak biomechanical loads when choosing and implementing a motor control program for balance recovery.

The prescribed short step length may be close to the smallest step length the young could utilize to successfully recover balance with a single step. Evidence for this is the mean landing time (460 ms) associated with the short step was comparable to the lower plateau of landing time observed in a previous study [21]. In addition, participants' short steps were consistently larger than the intended short step target position, indicating they had difficulty achieving the short step length (Fig. 4). This may help explain the marginal difference observed in balance recovery time. Although recovery time increased with increasing step length, the short step was consistently longer than planned, which may have reduced the difference in recovery time between the short and natural step tasks.

Our results may suggest insights into factors contributing to the multiple short step strategies preferred by older adults [6,7]. A short step reduces the biomechanical demands, which may be preferential for older adults who have reduced motor capacities (e.g. reduced muscle mass and contractile strength) compared to young [22]. Also, older adults, compared to young, have a reduced functional base of support [23,24]. Consequently the tendency of older adults to use multiple short steps may indicate reduced motor capacity since small steps are less biomechanically demanding. Further study may help determine the extent to which multiple step responses in older adults are the result of reduced physical capacity, cognitive (or pre-planned) issues, or are reactions to events occurring after the initiation of the first step [7].

The results of this study in conjunction with prior studies provide guidance for designing interventions that reprogram older adults' balance recovery strategies to reduce fall risk. For example, Hsiao and Robinovitch found that older adults, when taking multiple steps in response to a backwards lean-and-release perturbation, occasionally took such small steps that the BOS was unable to capture the backwards-moving COM [10]. Additionally, Judge et al. found that older adults' reduced plantarflexion strength capacity manifested itself as a reduction in step length. Since reduced step length is a common attribute among elderly fallers [10–12], increasing ankle plantarflexion strength and power are likely to decrease falling risk by increasing step length [25]. Since the results of the current study lend insight into the reduced strength requirements associated with small steps, programs that teach older adults to take larger steps [10] and emphasize increasing ankle plantarflexion strength [25] may be effective in lengthening older adults' natural steps and improving the ability to produce balance-restoring moments.

The main study limitation is that prescribing a specified step length may not result in a natural balance recovery mechanism. Although the perturbation magnitude used in this study was large enough to require a step to restore balance, the instructional constraint on foot placement invoked a voluntary mechanism whose biomechanical characteristics may differ from natural responses. For example, Varraine et al. observed that, when voluntarily modulating step length during treadmill walking, participants increased muscle activity during shorter and longer than natural steps and increased propulsive forces during longer than natural steps [14]. Similarly, prescribing a single step strategy may have forced an unnatural response for participants preferring a multiple step strategy, although this is unlikely since most young adults naturally use a single step strategy [6]. The characteristics of a single step are likely different from the first step of a multiple step response; therefore imposing a single step is useful to allow direct comparison of our results to previous and future experiments. A second study limitation is that only young participants were tested. Even so, our observation that young adults generate smaller AP force impulses during smaller steps is consistent with Won, who observed that older adults, compared to young, took smaller steps with smaller AP force impulses [20]. Although this demonstrates a connection between step length and force impulse across age groups, further testing on older participants is required to determine whether older adults possess the strength capacity to manage the larger AP force impulses associated with larger steps.

In summary, we found significant differences in balance parameters when using a shorter step for balance recovery. A shorter step resulted in an invariant step preparation phase followed by a variant recovery phase characterized by an earlier recovery time and a smaller AP impulse at landing. Given that the participants naturally chose not to use a step as short as they were capable of employing, the results support the idea that healthy young individuals do not prioritize minimizing recovery time nor strength requirements when selecting their step length.

Acknowledgments

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