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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].