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Short communication

Method for evoking a trip-like response using a treadmill-based perturbation during locomotion

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ABSTRACT

Because trip-related falls account for a significant proportion of falls by patients with amputations and older adults, the ability to repeatedly and reliably simulate a trip or evoke a trip-like response in a laboratory setting has potential utility as a tool to assess trip-related fall risk and as a training tool to reduce fall risk. This paper describes a treadmill-based method for delivering postural perturbations during locomotion to evoke a trip-like response and serve as a surrogate for an overground trip. Subjects walked at a normalized velocity in a Computer Assisted Rehabilitation Environment (CAREN). During single-limb stance, the treadmill belt speed was rapidly changed, thereby requiring the subject to perform a compensatory stepping response to avoid falling. Peak trunk flexion angle and peak trunk flexion velocity during the initial compensatory step following the perturbation were smaller for responses associated with recoveries compared to those associated with falls. These key fall prediction variables were consistent with the outcomes observed for laboratory-induced trips of older adults. This perturbation technique also demonstrated that this method of repeated but randomly delivered perturbations can evoke consistent, within-subject responses.

1. Introduction

Research on falls is increasingly important since falls are the leading cause of unintentional-injuries leading to death for the rapidly growing population of older adults in the United States (Centers for Disease Control and Prevention, 2013). As important, persons with lower limb amputation have a high incidence of falls. More than 50% of individuals with lower limb amputation fall annually (Miller et al., 2001), compared with a fall prevalence of 33% generally associated with older adults. Falls can lead to detrimental consequences such as loss of confidence, fear of falling, and injury. It is therefore important to find a way to assess the ability of these persons to recover from a large postural perturbation and find methods for training them to avoid a fall before injury occurs.

Only a handful of research studies have utilized novel methods to induce a laboratory trip during locomotion. These include an obstacle rising above the ground to obstruct the swing foot motion (Eng et al., 1994; Pavol et al., 1999; Pijnappels et al., 2004), restricting the swing foot motion using a cord or similar device (Blumentritt et al., 2009; Krasovsky et al., 2012; Smeesters et al., 2001), or dropping an obstacle on a treadmill to obstruct the swing leg forward movement (Schillings et al., 1996). These methods successfully induce a realistic trip, but repeated trips can be anticipated. In contrast, disturbances delivered by a computer-controlled treadmill system may offer an easily controlled and reproducible alternative. In this paper, we describe a method of delivering an unanticipated perturbation to subjects walking on a treadmill that evokes a repeatable recovery response that requires a compensatory stepping response to avoid falling.

2. Methods

Participants were active-duty members of the U.S. military with traumatic unilateral lower limb amputation who were recruited from the Comprehensive Combat and Complex Casualty Care program at the Naval Medical Center San Diego (NMCSD). In this study 12 male subjects (mean age: 24.3 ± 2.8 years, weight: 77.0 ± 14.3 kg, height: 177.5 ± 6.5 cm, and time since amputation: 25 ± 31 months) participated. Subjects were between 5 and 114 months post-amputation and highly functional (Medicare Functional Classification Level K3 or K4). Inclusion criteria were traumatic transtibial amputation, male, between the ages of 18 and 40 years

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Fig. 1. Subject walking within the Computer Assisted Rehabilitation Environment (CAREN) at the Naval Health Research Center.

who were ambulating without an assistive device, medically cleared for high-level functional activities, and could walk continuously for more than 15 min. Exclusion criteria for subjects included traumatic brain injury, vestibular dysfunction, and significant injury to the sound limb. The experimental protocol was approved by the NMCSD, Naval Health Research Center (NHRC), and Mayo Clinic institutional review boards as well as the Human Research Protection Office, U.S. Army Medical Research and Materiel Command. Subjects provided written informed consent prior to participating in the study.

Perturbations were delivered while subjects walked in a Computer Assisted Rehabilitation Environment (CAREN) extended version (Motek Medical BV, Amsterdam, The Netherlands). This immersive virtual environment consists of a six degrees of freedom motion platform (Moog Inc., East Aurora, New York) with a 1.7-m-long dual-belt (side-by-side) instrumented treadmill (Forcelink, B.V., Culemborg, The Netherlands) capable of high accelerations (up to 5 m/s^2). Visual inputs were synchronized with the subject's treadmill walking speed to simulate walking on an endless pastoral path (Fig. 1). The normalized walking speed for each subject was controlled for leg length and set at the Froude number (*Fr*) 0.2, where $Fr = v^2/gl$, v is the walking speed, g is the gravitational constant, and l is the leg length (Alexander, 1989). The selection of this speed was based on the overground self-selected walking speed of subjects with lower limb amputation previously studied at the NMCSD Motion Analysis Laboratory, and it was chosen to be slightly slower than self-selected walking velocity so that the subjects could maintain a constant speed for the duration of the perturbation trial (approximately 15 min) without fatigue. Walking speeds ranged from 1.0 to 1.5 m/s.

Each subject wore a full-body harness tethered to an instrumented safety system (Interface, Scottsdale, AZ) that could support the subject's full weight. The length of the tether was set so that, in the event of a fall, the subject's hands and knees would not contact the ground. The tether did not interfere with normal walking. A trial was categorized as a fall if more than 50% of the subject's body weight was supported by the safety harness (Brady et al., 2000). In addition, a trial was categorized as a fall if the rear (stance) foot triggered the rear safety light gate causing the treadmill belt motion to stop (treadmill length is 1.7 m). Trials where the harness supported 20–50% of body weight were classified as harness assisted and were not included in the analyses. In addition, trials during which the stepping response caused the subject to step sideways off of the treadmill were excluded from the analysis.

Each subject first walked for 10 min at the normalized walking speed in order to become acquainted with treadmill locomotion. This warm-up period was followed by a 6-min walk at the same speed during which six perturbations, three each for the prosthetic side and sound side limbs, were delivered at random times to evoke a trip-like response. Initiation of the perturbation profile was triggered when initial contact of the selected foot and a force of 40 N was detected by the underlying force plate. The perturbation profile was defined by a series of changes in treadmill velocity (Fig. 2) and began with the treadmill decelerating (-15 m/s^2) from the normalized walking velocity for 50 ms, followed by a treadmill acceleration (15 m/s^2) for 270 ms. This temporarily brought the treadmill to a velocity between 4.3 and 4.8 m/s (depending on the subject's normalized walking speed) before decelerating (-15 m/s^2) back to the normalized walking speed. Both belts moved at the same speed throughout the perturbation. The elapsed time of the perturbation was less than a second. The timing of these changes in velocity was chosen so that, during the second phase of the perturbation (treadmill acceleration) the foot selected to be perturbed was in stance and the contralateral limb was in swing (usually at midswing half way through phase 2). The initial deceleration phase helped to arrest the velocity of the stance foot while allowing the trunk to continue forward, thereby increasing the dynamic stability margin (Hof et al., 2005) and making the recovery response during the acceleration phase more challenging. The limb that was on the ground (stance limb) during the treadmill acceleration is described as the perturbed limb in this paper.

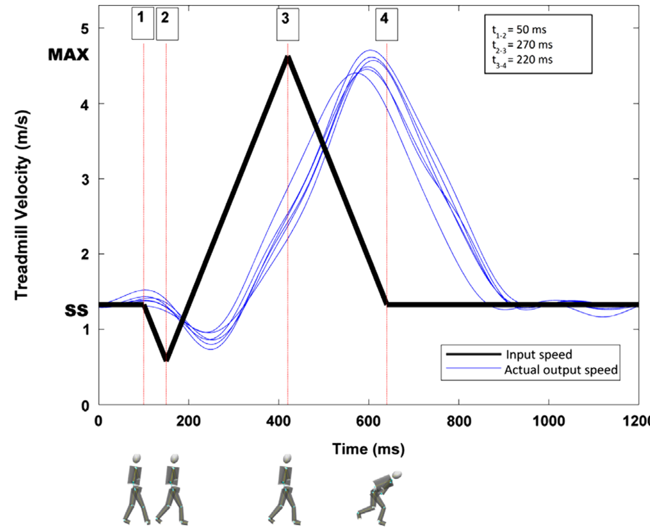

Fig. 2. Perturbation timing and velocity profile utilized to perturb subjects on the treadmill during gait. The four vertical time points mark where a change in the treadmill velocity occurred and figures below the graph show the approximate position of the subject at this time point. Thick line indicates the input speed sent to the treadmill. Thinner lines illustrate the actual output speed of the treadmill for several trials, showing the variability of the signal. SS is the steady state speed, which ranged from 1.0 to 1.5 m/s for the participating subjects.

The subject was instructed to recover from the perturbation as best he could and continue walking if possible. If the subject fell, the treadmill was stopped and the subject was allowed to rest, if desired, before continuing the protocol. If the subject was able to fully recover, he continued walking at the normalized velocity until another perturbation was delivered or data collection was complete.

For the study, 34 retroreflective markers were placed on the subject using a modified Helen Hayes marker set configuration (Kadaba et al., 1990). The motion of the markers was tracked using a 12-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) operating at 120 Hz. Marker data were filtered using a fourth-order bidirectional recursive Butterworth filter with a cutoff frequency of 9 Hz in Visual3D (C-Motion, Inc., Germantown, MD, USA). A 13-segment rigid body model using the marker data was created to represent the whole body. Trunk flexion angle, defined as the angle of the trunk segment with respect to vertical, was calculated from time of the perturbation to initial recovery step. Trunk flexion velocity was computed as the derivative of the trunk flexion angle. Root mean square (RMS) error of trunk flexion angle and velocity were calculated for the second and third trips on each side.

A one way ANOVA was used to compare peak trunk flexion angle and peak trunk flexion velocity during the perturbation between trials that were classified as a fall to those outcomes that were classified as a recovery on the prosthetic limb (no falls occurred during the sound limb trials). A one way ANOVA was also run to compare the difference in peak trunk flexion angle and velocity between the sound and prosthetic limb recovery (non-fall) trials. Statistical tests were performed using SPSS 18 (IBM Corporation, Armonk, NY, USA) and statistical significance was set at $p \le 0.05$.

3. Results

The perturbation protocol was successfully completed by 12 subjects. Four of the 12 subjects experienced a fall during the induced perturbation, with the fall occurring during the initial perturbation on the prosthetic side limb. Two of these four subjects fell again during the second perturbation on the same side. Three subjects experienced a harness assist trial, two on the prosthetic side limb and one on the sound side limb. In total, of the 72 cumulative perturbations from all the subjects, 63 trials were classified as a successful recovery, three were classified as a harness assist, and six were classified as a fall, with all of the falls occurring when the prosthetic side limb was perturbed.

For perturbations on the prosthetic side limb (i.e. the prosthetic limb is on the ground when the treadmill perturbation begins), mean peak trunk flexion angles associated with recoveries ($31 \pm 12°$) were