Study Title: Locomotor Response of Persons with Upper Limb Loss to Treadmill Perturbations

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Background and Study Design

The role of the upper limbs in gait dynamics has received little attention historically, however surmounting evidence supports an important function of the arms in balance during locomotion and other dynamic tasks. Regaining balance after a perturbation is contingent on reactive mechanisms for re-stabilization. Rapid arm elevation is commonly observed in able-bodied individuals following a perturbation, which acts to increase the moment of inertia (I), slowing trunk motion. In any movement context, it is plausible that the result of a perturbation, i.e., a fall or recovery, could rest on the ability to effectively manipulate arm motion. A high fall rate has been observed in individuals with upper limb absence (ULA). A large proportion of falls in persons with ULA (67%) has been reported to occur during walking activities. Effects of ULA-induced imbalance can be observed in an increase in sound arm swing during walking. Use of an upper limb prosthesis, which effectively reduces the mass discrepancy, might be a solution to this issue of imbalance. Reactive responses are yet to be explored in individuals with ULA with and without use of a prosthesis. The objective of this cross-over intervention study was to investigate the reactive responses of people with ULA to perturbations during walking.

Protocol

Participants completed two baseline walking trials and 12 perturbation trials on an instrumented treadmill (Motek, Enschede, Netherlands). Baseline trials consisted of 30 seconds of steady-state walking, first at a self-selected speed then 1.0 m/s. The perturbation was designed to disrupt walking and instigate trip-like responses: the treadmill belt was programmed to accelerate (6.5 m/s^2) to 3.3 m/s, then immediately decelerate (6.5 m/s^2) to 1.0 m/s. During each trial a perturbation was triggered at initial contact of a randomly-selected step between 21 and 40 after the belt reached 1.0 m/s and delivered during single limb stance of the sound/dominant or impaired/non-dominant arm side (6 per side). Participants walked for twenty strides after restoration of reciprocal gait before the treadmill was stopped. Side/step randomization minimized anticipatory effects of knowledge of the perturbation timing on motor response. Prosthesis users completed all tasks with and without their customary prosthesis, order randomized.

Mass, COM location and *I* of each prosthesis were estimated using an oscillation cage and reaction board. *I* was estimated about the medial-lateral axis of rotation at the prosthesis proximal end and translated to its COM using the parallel axis theorem.

Data Analysis

Kinematic data were collected with a 12-camera motion capture system (Motion Analysis, Rohnert Park, CA) at 120 Hz. Data were exported to Visual 3D (C-motion, Germantown, MD), filtered using a 4th-order low-pass (9 Hz) Butterworth filter, and a customized 12-segment model was applied. The assumption of uniform density [36] was applied to estimate segment mass as a percentage of body mass, COM position and *I*, except for the ULA impaired limb where the forearm was modified based on the prosthesis condition.

Initial contact events were labelled using a velocity-based algorithm and corrected manually. L was estimated by calculating the instantaneous angular momentum of each segment about the COM in three planes, summing the values according to Eq. 1, where, for each segment i, $\overrightarrow{\omega_i}$ is the angular velocity, $\vec{r_i}$ is the distance from segment COM to whole-body COM, m_i is the mass and $\vec{v_i}$ is relative velocity.

$$\vec{L} = \sum_{i=1}^{n} \vec{I_i}\overrightarrow{\omega_i} + \vec{r_i} \times m_i\vec{v_i} \qquad \text{Eq. 1}$$

L was normalized by treadmill speed and participant mass and height. L_{range} was calculated as maximum minus minimum L over the period of interest. For baseline trials, L_{range} was calculated over each stride and averaged over the middle 10 strides. The perturbation recovery phase was defined as perturbation onset time to the seventh initial contact, and L_{range} was calculated from global maximum and minimum across that period. To aid interpretation of L_{range}, bilateral shoulder flexion-extension and add-abduction ranges were calculated similarly for each period of interest. Shoulder flexion-extension was defined as the sagittal-plane angle between the upper arm long axis (line connecting shoulder and elbow markers) and thorax long axis (line connecting midpoint between shoulder markers and pelvis center) and add-abduction angle was that relative angle in the coronal-plane. Five perturbation trials on each side were averaged with the first (naïve) trial excluded because of exaggerated reactions to first-side perturbation.

Statistical Analysis

Descriptive statistics are produced for L_{range} (recovery minus baseline), shoulder joint range-of-motion during recovery, maximum trunk flexion angle, and maximum trunk flexion velocity for each condition (controls: with arms free and with non-dominant arm bound; upper limb absence: without wearing the prosthesis and with wearing the prosthesis).