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Control of locomotor stability in stabilizing and destabilizing environments

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ABSTRACT

To develop effective interventions targeting locomotor stability, it is crucial to understand how people control and modify gait in response to changes in stabilization requirements. Our purpose was to examine how individuals with and without incomplete spinal cord injury (iSCI) control lateral stability in haptic walking environments that increase or decrease stabilization demands. We hypothesized that people would adapt to walking in a predictable, stabilizing viscous force field and unpredictable destabilizing force field by increasing and decreasing feedforward control of lateral stability, respectively. Adaptations in feedforward control were measured using after-effects when fields were removed. Both groups significantly ($p < 0.05$) decreased step width in the stabilizing field. When the stabilizing field was removed, narrower steps persisted in both groups and subjects with iSCI significantly increased movement variability ($p < 0.05$). The after-effect of walking in the stabilizing field was a suppression of ongoing general stabilization mechanisms. In the destabilizing field, subjects with iSCI took faster steps and increased lateral margins of stability ($p < 0.05$). Step frequency increases persisted when the destabilizing field was removed ($p < 0.05$), suggesting that subjects with iSCI made feedforward adaptions to increase control of lateral stability. In contrast, in the destabilizing field, non-impaired subjects increased movement variability ($p < 0.05$) and did not change step width, step frequency, or lateral margin of stability ($p > 0.05$). When the destabilizing field was removed, increases in movement variability persisted ($p < 0.05$), suggesting that non-impaired subjects made feedforward decreases in resistance to perturbations.

1. Introduction

Walking-intensive interventions have a high probability of improving walking speed of individuals with motor incomplete spinal cord injury (iSCI) [1,2]. However, dynamic balance deficits remain a significant problem [3–5] as 75% of ambulatory individuals with iSCI fall each year [3]. We need better methods to enhance gait stability [6], the ability to recover from perturbations. To address this deficit, gait training often includes balance-challenging tasks [7] and/or stability assistance [8]. In specific contexts, contrasting intervention tools of perturbation training [9,10] and kinematic assistance [8] can each improve balance. How best to integrate these techniques into current practice remains unclear.

People use feedforward and feedback mechanisms to control rhythmic movements [11]. Feedforward strategies include internal models and impedance mechanisms that are particularly valuable for responding to predictable and unexpected disturbances, respectively [12]. With neurologic impairment, reliance on impedance mechanisms (e.g. posture and muscular co-contractions [13–15]) to resist perturbations can compensate for decreased ability to use feedback mechanisms (e.g. corrective steps), which require accurate sensing of and response to stimuli. Following iSCI, cautious gait patterns, including wide steps [16] and increased double-support time [5], suggest that impedance mechanisms are utilized. In contrast, non-impaired populations likely minimize impedance contributions to gait stability due to negative impacts on energetic efficiency [16–18] and maneuverability [19] during community ambulation.

We observed locomotor adaption in stabilizing and destabilizing environments and quantified the presence of after-effects (indicative of feedforward adaptations) to gain insight into neural control mechanisms. We examined ambulatory individuals with iSCI and also non-impaired participants to better understand how sensory-motor function impacts stabilization strategy.

Given that people optimize locomotion for effort and error [20,21], we hypothesized that when provided external lateral stabilization that reduces movement errors, participants would reduce feedforward

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mechanisms to control frontal-plane momentum, selecting narrower, less metabolically-costly steps instead [18]. Additionally, we theorized that participants would exhibit after-effects of increased movement variability and decreased lateral margins of stability (MOS) when external stabilization was removed. For individuals with iSCI, suppression of feedforward stabilization mechanisms could result in instability due to limited ability to offset this adaptation with feedback mechanisms. Conversely, we hypothesized that when challenged with unpredictable perturbations, participants would control frontal plane movements by adapting feedforward stability mechanisms, including wider, faster steps and increased lateral MOS [22,23]. We anticipated that both subject groups would show after-effects of increased stability (increased MOS and step width) when the perturbations ceased. Understanding how people adapt to stabilization assistance and perturbations will be valuable for effective integration of these contrasting intervention tools into programs targeting enhancement of gait stability.

2. Materials and methods

2.1. Subjects

A convenience sample of 8 ambulatory participants with chronic motor incomplete iSCI (AIS D) and 10 non-impaired subjects gave written informed consent. One subject with iSCI could not complete all experimental conditions and was excluded from analysis. Northwestern University Institutional Review Board approved the protocol. Participants with iSCI were 58 ± 8 years, weight 73 ± 38 kg, and 6 males/1 female (Supplement Table 1 for clinical outcome measures). Non-impaired participants were 24 ± 4 years, weight 69 ± 8 kg, and 6 males/4 females. See Supplement Appendix A for detailed inclusion/exclusion criteria.

2.2. Experimental setup

Subjects walked on an oversized treadmill, belt width 1.39 m (Tuff Tread, Willis, TX), providing space to respond to perturbations (Fig. 1a). Subjects wore a harness attached to a passive overhead safety device (Aretech, Ashburn, VA) that provided no bodyweight support.

Lateral forces were applied using two cables attached to a pelvis harness (Fig. 1a). A series-elastic linear motor created force on each cable. Load cells measured the forces. Applied forces varied with experimental condition. During the *Stabilization* condition, subjects experienced a variable force proportional in magnitude and opposite in direction to real-time lateral center of mass (COM) velocity; viscosity gains were 427 ± 78 N/(m/s) and applied forces did not exceed 110 N. This viscous field reduced the requirements to actively maintain straight-ahead walking. During the *Destabilization* condition, random bidirectional force perturbations normally distributed from −33 to 33 N were applied at 3 Hz. Perturbation magnitude was selected to be challenging but manageable for participants with iSCI. Perturbation frequency was faster than step frequency to encourage feedforward adaptations. Perturbations increased requirements to actively maintain straight-ahead walking.

A 10-camera motion capture system (Qualysis, Gothenburg Sweden) recorded 3D coordinates of reflective markers located on the pelvis (superior iliac crests, anterior-superior iliac spines, S2, and 2 tracking markers) and bilaterally on the greater trochanter, lateral knee, lateral malleolus, calcaneous, and second and fifth metatarsals during gait.

2.3. Protocol

First, we collected demographic (both groups) and clinical outcome measures (iSCI only) at preferred overground walking speeds without assistive devices. Next, we identified preferred treadmill walking speed (non-impaired 1.2 ± 0.1 m/s; iSCI 0.3 ± 0.2 m/s). Subjects were instructed to walk as they felt most comfortable, swing their arms freely, and keep their midline centered over a line drawn along the treadmill center. No handrails or assistive devices were used.

Then subjects walked at preferred speed during three lateral force conditions (Fig. 1b). During each condition, subjects completed 400 steps. The first 100 steps established a **Baseline** measure of walking with no external assistance. The treadmill was then stopped, and a force **Field** was applied during standing. The force Field conditions were; 1) *Stabilization* – viscous lateral force field, 2) *Destabilization* – random lateral force perturbations, or 3) *Null* – no applied forces. The treadmill was restarted, and the next 200 steps occurred in the force **Field**. Any applied forces were then removed without stopping the treadmill, and subjects walked another 100 steps to measure any **After-effects**. The condition order was randomized. Subjects with iSCI rested at least 2 min between conditions.

During the Stabilization and Destabilization conditions, subjects received a verbal countdown 5 steps before the applied forces were removed. When no forces were applied (Baseline, After-effects, and Null Field) the cables remained attached to the pelvic harness but hung slack.

2.4. Data processing

Kinematic marker data was processed using Visual3D (C-Motion, Germantown, MD) and a custom MATLAB (Mathworks, Natick, MA) program. Marker data was low-pass filtered (Butterworth, 6 Hz cut-off frequency) and gap-filled. Time of initial foot contact (IC) and toe-off (TO) were identified at each step based on fore-aft positions of the calcaneous and 5th metatarsal markers. A Visual 3D pelvis model was created using the 7 pelvis markers. Mediolateral COM position was calculated as the center of the pelvis model.

For each step we identified peak lateral COM speed as a net measure of COM control. To assess how control was instituted, we calculated step width, step time, and minimum MOS [24]. Step width was calculated as the medio-lateral distance between the left and right 5th Metatarsal markers at IC. COM velocity was calculated as the derivative of COM position. Peak lateral COM speed was identified as the maximum absolute COM velocity between IC events. Step time was calculated as time between successive IC's.

MOS was calculated using the following equation [25] to first identify the extrapolated center of mass (XCOM) position:

$$XCOM = COM + COM'*\sqrt{l/g}$$

$XCOM = lateral\ extrapolated\ center\ of\ mass$

$COM = lateral\ center\ of\ mass\ position$

$COM' = lateral\ center\ of\ mass\ velocity$

$l = pendulum\ length$

$g = gravitational\ constant$

"*l*" was calculated as the instantaneous distance between the COM and the lateral malleolus.

MOS was calculated as the distance between the XCOM and the base of support (BOS), approximated as the lateral position of the 5th metatarsal marker on the side of the last IC. MOS was positive when the XCOM was medial of the BOS. Minimum MOS was identified during stance phase of each step.

To estimate the time course of any after-effects, we fit an exponential function [26] to all kinematic metrics, but step width was most robust at describing the observed After-effects period (Fig. 2a). Exponential fits were only consistently significant (linear regression F-test,