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Decreased lower limb muscle recruitment contributes to the inability of older adults to recover with a single step following a forward loss of balance

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ABSTRACT

In response to a balance disturbance, older individuals often require multiple steps to prevent a fall. Reliance on multiple steps to recover balance is predictive of a future fall, so studies should determine the mechanisms underlying differences between older adults who can and cannot recover balance with a single step. This study compared neural activation parameters of the major leg muscles during balance recovery from a sudden forward loss of balance in older individuals capable of recovering with a single step and those who required multiple steps to regain balance. Eighty-one healthy, community dwelling adults aged 70 ± 3 participated. Loss of balance was induced by releasing participants from a static forward lean. Participants performed four trials at three initial lean magnitudes and were subsequently classified as single or multiple steppers. Although step length was shorter in multiple compared to single steppers ($F = 9.64$; $p = 0.02$), no significant differences were found between groups in EMG onset time in the step limb muscles ($F = 0.033–0.769$; $p = 0.478–0.967$). However, peak EMG normalised to values obtained during maximal voluntary contraction was significantly higher in single steppers in 6 of the 7 stepping limb muscles ($F = 1.054–4.167$; $p = 0.045–0.024$). These data suggest that compared to multiple steppers, single steppers recruit a larger proportion of the available motor unit pool during balance recovery. Thus, modulation of EMG amplitude plays a larger role in balance recovery than EMG timing in this context.

1. Introduction

In response to a balance disturbance in humans, a combination of neural, biomechanical and environmental factors influence the ability to recover balance. Tripping represents one of the most common balance disturbing stimuli, and experimental approaches have been designed with the aim of replicating the stepping response to a trip. A common approach is the tether-release method, whereby subjects are leaned forward whilst suspended from a horizontal cable attached to their trunk. The cable is subsequently released after a random time delay and the individual experiences a forward loss of balance that requires a rapid forward step to prevent the occurrence of a fall (Do et al., 1982; Thelen et al., 1997). Older individuals, in contrast to their younger counterparts, often require multiple steps to arrest their forward and inferior momentum and subsequently avoid contact with the ground, although some older individuals are capable of recovering their balance with a single step (Carty et al., 2012c; Thelen et al., 2000).

The reliance on multiple steps to recover from a forward loss of balance is predictive of a future fall (Maki et al., 2001), so it is important to determine the mechanisms underlying differences between older adults who can recover balance with a single step compared to those who cannot. To date, assessment of differences in balance recovery using the tether-release method between single and multiple steppers have focused on kinematic and kinetic measures (Arampatzis et al., 2008; Barrett et al., 2012; Carty et al., 2012b; Carty et al., 2012c; Carty et al., 2011; Karamanidis et al., 2008; Madigan, 2006; Madigan and Lloyd, 2005; Wojcik et al., 1999, 2001), but potential differences in neural parameters have not been examined.

Among older adults, a larger first step decreases the likelihood that multiple steps are required to regain balance (Carty et al., 2012c) because the whole body centre of mass remains further behind the anterior boundary of the base of support at foot contact (Karamanidis et al., 2008; Carty et al., 2012c). The ability to rapidly

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move the stepping limb (Carty et al., 2012b; Madigan, 2006; Thelen et al., 1997; van Dieen et al., 2005), generalised lower extremity muscle weakness (Carty et al., 2012a; Grabiner et al., 2005; Pijnappels et al., 2008a,b) and differences in muscle activation timing (Thelen et al., 2000) have been proposed as important determinants of stepping strategy. It might therefore be expected that differences in neural activation patterns are evident between single and multiple steppers. The aim of this study was to compare the timing and magnitude of activation of the major leg muscles during balance recovery from a sudden forward loss of balance in older individuals capable of recovering with a single step and those who required multiple steps to regain balance. We hypothesised that multiple steppers would exhibit reduced muscle activation magnitudes and delayed onset times compared to single steppers.

2. Methods

2.1. Participants

Eighty-one healthy, community dwelling older adults (age: 70 ± 3 years; height: 1.66 ± 0.09 m; body mass: 75.8 ± 12.7 kg; sex: 45 males, 36 females) were randomly recruited from the electoral roll to participate in the study. Individuals who reported neurological, cognitive, metabolic, cardiovascular, pulmonary or musculoskeletal impairment were excluded. All participants had normal or corrected to normal vision. The study was undertaken in accordance with the Declaration of Helsinki. Ethical approval was obtained from the Institutional Human Research Ethics Committee and all participants gave informed consent.

2.2. Balance recovery assessment

The balance recovery protocol was conducted as described previously (Carty et al., 2012c). Participants stood barefoot with their feet shoulder-width apart in an neutral posture and were tilted forward, keeping their feet flat on the ground, until 15%, 20% or 25% of body weight (BW) was recorded on a load cell (S1W1kN, XTRAN, Australia) placed in series with an inextensible cable. One end of the cable was attached to a safety harness worn by the participant at the level of their sacrum and the other end was attached to a rigid metal frame located behind the participant. An electric winch, mounted on the frame, was used to adjust the length of the cable. Care was taken to ensure the cable was aligned parallel with the ground and that participants kept their head, trunk and extremities aligned prior to cable release. The cable was released at a random time interval (2–10 s) following achievement of the prescribed posture and cable force (±1% BW), through the disengagement of an electromagnet located in-series with the cable. No differences in cable force or release angle (computed from the sagittal plane angle between the vertical, and a line connecting the ankle joint centre with the whole body centre-of-mass) were detected at cable release between single and multiple steppers for any lean magnitude ($p > 0.05$). Participants were instructed to relax their muscles while leaning and to regain balance with a single step using the stepping lower limb of their choice, once they perceived that they were falling. The instruction to take a single step was reiterated prior to every subsequent trial. A second cable, instrumented with a load cell (S1W1kN, XTRAN, Australia), attached the safety harness to the ceiling, and was used to prevent participants from contacting the ground in the event of a fall. Ground reaction forces were acquired at 1 kHz using a single piezoelectric force platform (Type 9287A, Kistler Instrument Corporation, USA) located under both feet in the pre-release position. Overhead cable force and centre of pressure location were displayed in real time on a computer monitor and were visually inspected by the investigator to ensure anticipatory actions (e.g., antero-posterior and medio-lateral weight shifting) were not evident in the period immediately prior to release. Following an initial practice trial at 15% BW lean magnitude, participants performed 4 trials at each lean magnitude, with block randomisation used to determine the lean magnitude sequence (i.e., 15%, 20% or 25% BW) for the 12 trials, however only results from the 20% BW condition were considered in this report.

2.3. Biomechanical analysis

Trajectories of 51 reflective markers attached to the head, trunk, pelvis, and upper and lower limbs were recorded at 200 Hz using a 10-camera, three-dimensional motion capture system (Vicon Motion Systems, Oxford, UK). Ground reaction forces for the stance and stepping feet were simultaneously acquired at 1 kHz using two 900×600 mm piezoelectric force platforms (Kistler, Amherst, USA). Whole body three-dimensional balance recovery kinematics were computed as described previously (Barrett et al. 2012). Three events in the balance recovery task were identified: (1) cable release, defined as a 20% reduction in force measured using a force transducer in series with the restraining cable (2) Toe off, determined from toe marker kinematics (De Witt, 2010) and (3) foot contact, defined as a force in excess of 5 N recorded on the anterior force plate. Step length was calculated as the horizontal distance from the great toe marker on the rear leg to the corresponding marker on the step leg.

2.4. EMG analysis

Surface electromyography (EMG) activity was recorded using bipolar surface electrodes (Duo-trode, Myotronics Inc., Australia) positioned along muscle fibre direction with an inter-electrode distance of 2 cm. Skin preparation included shaving of hair, and skin abrasion with abrasive alcohol wipes. Data were collected telemetrically (Aurion ZeroWire; Milano, Italy) from 7 muscles of each leg: rectus femoris, vastus medialis, semitendinosus, biceps femoris, medial gastrocnemius, soleus and tibialis anterior at 1 kHz. To enable normalisation of the EMG signals, maximal voluntary contractions were performed with the ankle dorsi- and plantar flexors (0° plantar flexion) and the knee flexors and extensors (90° of knee flexion) in a Biodex dynamometer (Biodex Medical Systems Inc., USA). Two of each contraction type were performed with at least 1 min rest periods between contractions. To detect muscle activity onsets, the raw EMG traces were band-pass filtered at 10–500 Hz, full-wave rectified and then low-pass filtered at 100 Hz using a 2nd order zero phase-shift Butterworth filter to preserve the original onset of activity. Mean EMG activity was determined from the 500 ms period preceding the start of the trial. Onset latency was then defined as the time between tether release and the instant that EMG amplitude exceeded 3 standard deviations of the mean pre-release EMG for at least 25 ms. To normalise muscle activity during the balance recovery task, EMG signals were root mean square integrated and low-pass filtered at 4 Hz. EMG data during the stepping movement were then expressed relative to the mean integrated EMG during the 1s period surrounding the maximum torque phase of the respective MVC.

2.5. Participant classification

Participants were classified as either single steppers (a single step across all four trials at a given lean magnitude), multiple steppers (multiple steps across all four trials at a given lean magnitude) or mixed steppers (a combination of single and multiple steps across trials at a given lean magnitude). The criteria used to distinguish a multiple versus single step recovery strategy were: (1) a