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Muscle coordination of support, progression and balance during stair ambulation

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ABSTRACT

Stair ambulation is more physically demanding than level walking because it requires the lower-limb muscles to generate greater net joint moments. Although lower-limb joint kinematics and kinetics during stair ambulation have been extensively studied, relatively little is known about how the lower-limb muscles accelerate the whole-body center of mass (COM) during stair ascent and descent. The aim of the current study was to evaluate differences in muscle contributions to COM accelerations between level walking and stair ambulation in 15 healthy adults. Three-dimensional quantitative gait analysis and musculoskeletal modeling were used to calculate the contributions of the individual lower-limb muscles to the vertical, fore-aft and mediolateral accelerations of the COM (support, progression, and balance, respectively) during level walking, stair ascent and stair descent. Muscles that contribute most significantly to the acceleration of the COM during level walking (hip, knee, and ankle extensors) also dominate during stair ambulation, but with noticeable differences in coordination. In stair ascent, gluteus maximus accelerates the body forward during the first half of stance and soleus accelerates the body backward during the second half of stance, opposite to the functions displayed by these muscles in level walking. In stair descent, vasti generates backward and medial accelerations of the COM during the second half of stance, whereas it contributes minimally during this period in level walking. Gluteus medius performs similarly in controlling mediolateral balance during level walking and stair ambulation. Differences in lower-limb muscular coordination exist between stair ambulation and level walking, and our results have implications for interventions aimed at preventing stair-related falls.

1. Introduction

Stair ambulation is a common activity of daily living. Although healthy adults can perform this task with relative ease, ascending and descending stairs can be more demanding for people with compromised motor function, such as the elderly (Reeves et al., 2008) or individuals with osteoarthritis (Kaufman et al., 2001; Asay et al., 2009). Compared to level walking, stair ambulation is associated with greater risk of severe or fatal falls (Manning, 1983), where 75% of these falls occur during stair descent (Svanström, 1974; Tinetti et al., 1988). Since muscles are responsible for controlling body movement, a better understanding of how muscles accelerate the whole-body center of mass (COM) (henceforth referred to as muscle function) during stair ambulation could help facilitate the development of more effective falls prevention strategies.

Lower-limb muscle function during level walking has been extensively investigated using musculoskeletal modelling approaches. Each muscle contributes to the vertical, fore-aft and mediolateral accelerations of the COM during stance (described as support, progression, and balance, respectively) (Pandy and Andriacchi, 2010). Liu et al. (2006) and Pandy et al. (2010) reported that gluteus medius, gluteus maximus, vasti, and soleus contribute significantly to support in the first half of stance, whereas forward progression in the second half of stance is dominated by soleus and gastrocnemius. Furthermore, to maintain balance in the frontal plane, Pandy et al. (2010) and John et al. (2012) showed that gluteus medius coordinates with vasti in the first half of stance while gluteus medius coordinates with both soleus and gastrocnemius in the second half of stance. By comparison, less is known about how the lower-limb muscles coordinate motion of the COM during stair ambulation.

Inverse dynamics-based studies suggest that greater knee and ankle extension moments are exerted during the first half of stance in stair ambulation than during this period in level walking (Riener et al., 2002; Silverman et al., 2014). These studies have also shown the peak knee extension moment during the second half of stance in stair descent to be more than three-fold greater than that observed during level walking. Finally, the ankle plantarflexion moment can peak as high as 75% of a maximal voluntary contraction in the early stance phase of stair descent (Reeves et al., 2008),

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whereas the ankle plantarflexion moment peaks during late stance in level walking.

Inverse dynamics-based studies have also investigated differences in the hip joint moment between level walking and stair ambulation. Compared to level walking, Riener et al. (2002) and Silverman et al. (2014) found the peak hip extension moment in the early stance phase of stair ascent and descent to be significantly smaller, with a larger reduction evident during stair descent. A few studies have compared the peak hip abduction moment between level walking and stair ambulation and have reported some inconsistent results. For example, Silverman et al. (2014) found the first and second peaks of the hip abduction moment during stair ascent to be significantly lower than those measured for level walking, whereas Nadeau et al. (2003) found no significant difference in the magnitude of the first peak.

While the aforementioned studies have provided important insights into the differences in net joint moments between level walking and stair ambulation, the corresponding changes in the functional roles of the individual lower-limb muscles can only be inferred from these differences (Zajac and Gordon, 1989). The reported differences in the magnitudes and/or timing of the lower-limb joint moments between level walking and stair ambulation suggest that there may also be differences between these two activities in the way the hip, knee, and ankle extensor muscles coordinate motion of the COM.

In the present study, we used a three-dimensional musculoskeletal model to investigate how lower-limb muscle function during stair ambulation differs from that during level walking. We anticipated that any differences in muscle contributions to COM motion will most likely be evident in the vertical direction because of the roles of the hip, knee, and ankle extension moments in supporting the body (Kepple et al., 1997) and the need to control the vertical COM displacement during stair ambulation. Given that stair ambulation has been demonstrated to be associated with greater knee and ankle extension moments but a reduced hip extension moment compared to level walking, the vertical support provided by these three extension moments should also vary accordingly. We therefore hypothesized that during stair ambulation the peak contributions to the vertical acceleration of the COM would be significantly increased for the knee and ankle extensors but significantly reduced for the hip extensors. The results of this study are expected to provide new insights into which lower-limb muscles are most relied upon for support, progression and balance during stair ambulation, and thus likely play a pivotal role in preventing stair-related falls.

2. Methods

Fifteen healthy adults (4 males, 11 females; age: 54 ± 8 yrs; weight: 67 ± 11 kg; height: 166 ± 8 cm) underwent gait experiments in the Biomotion Laboratory at the University of Melbourne. Ethical approval was obtained from the University of Melbourne Human Research Ethics Committee, and each participant provided written informed consent prior to the commencement of the study. Reflective markers were placed at specific anatomical landmarks on the trunk, pelvis, and both arms and legs. Marker trajectories were captured at 120 Hz using a nine-camera motion capture system (Vicon, Oxford Metrics Ltd, Oxford, UK) during all locomotor tasks. Pairs of Ag/AgCl surface electrodes (MediMax Global, Shalden, Hampshire, UK) were placed on an arbitrarily chosen leg to record the electromyographic (EMG) signal from five muscles: gluteus maximus, gluteus medius, vastus lateralis, gastrocnemius, and soleus. Additional details regarding retro-reflective marker and EMG electrode placement have been reported previously (Crossley et al., 2012). Ground reaction forces (GRFs) during gait were measured using a series of three ground-embedded force plates (Advanced Mechanical Technology Inc., Watertown, MA, USA), whereas GRFs during stair ambulation were measured using one ground-embedded force plate and two portable AccuGait force plates (Advanced Mechanical Technology Inc., Watertown, MA, USA) mounted on the first and second steps of a custom-built three-step staircase. GRF and EMG data were sampled at 1080 Hz.

All participants performed level walking (1.36 ± 0.15 m/s), stair ascent (0.50 ± 0.11 m/s) and stair descent (0.74 ± 0.20 m/s) tasks at a self-selected speed while wearing standardized footwear. Participants were asked to stand still in their neutral pose before performing any task. They were then instructed to land their test leg on the second ground-embedded force plate and the first step of the staircase during level walking and stair ambulation, respectively. Each trial commenced from initial contact with the test leg, and only data for the stance phase were analyzed. EMG data were also collected whilst all participants performed isometric maximum voluntary contractions of the muscles crossing the hip, knee and ankle. Marker and GRF data were low-pass filtered at 4 and 60 Hz, respectively, using a fourth-order Butterworth filter. EMG data were full-wave rectified and low-pass filtered at 10 Hz using a second-order Butterworth filter to create linear envelopes, which were normalized by the mean EMG signals recorded from each subject's maximum voluntary contraction trials.

A generic three-dimensional musculoskeletal model was implemented in an open-source software package (Delp et al., 2007) to calculate joint kinematics, joint kinetics and muscle forces based on the experimental data. The skeleton was represented as a 12-segment, 23 degree-of-freedom linkage system. The head and trunk were modelled as a single rigid body that articulated with the pelvis via a ball-and-socket joint. For the lower limbs, each hip was modelled as a ball-and-socket joint, each knee as a translating hinge joint, and each ankle as a universal joint comprised of two non-intersecting hinge joints. The lower limbs and trunk were actuated by 92 muscle-tendon units, with each unit represented as a three-element Hill-type muscle in series with an elastic tendon (Zajac, 1989). For the upper limbs, each shoulder was modelled as a ball-and-socket joint and each elbow was represented as a universal joint comprised of two non-intersecting hinge joints. The joints of the upper limbs were actuated by ten ideal torque motors (Dorn et al., 2012).

Scaled-generic models were developed by scaling the segmental inertial properties and muscle-tendon attachment sites assumed in the generic musculoskeletal model to each participant's body dimensions. Joint angles were computed over an entire gait cycle using an inverse kinematics analysis that minimized the sum of the squared differences between the positions of virtual markers identified on the model and reflective markers placed on the subject (Lu and O'Connor, 1999). Internal joint moments were calculated using a standard inverse dynamics approach.

Joint moments were decomposed into individual muscle forces using a static optimization algorithm, which minimized the sum of all muscle activations squared subject to each muscle's force–length–velocity properties (Anderson and Pandy, 2001). A pseudo-inverse force decomposition method (Lin et al., 2011) was then used to compute the contributions of all lower-limb muscle forces to the vertical, fore-aft, and mediolateral accelerations of the COM (support, progression, and balance, respectively). Individual muscle forces, as well as their contributions to the COM accelerations, were combined into functional muscle groups (see Fig. 2 caption). All results were time-normalized to the stance phase and then averaged separately across all participants. Muscle forces and joint moments were normalized to each participant's body weight and to body weight multiplied by height, respectively.

One-way repeated-measures ANOVA tests were used to determine whether locomotor task (i.e., level walking, stair ascent, and stair descent) significantly influenced the peak muscle forces and peak muscle contributions to the COM accelerations. If a significant main effect was obtained, post hoc paired *t*-tests were used to determine if significant differences existed between each of the locomotor tasks. A significance level of $p < 0.017$ was set for all tests after applying a Bonferroni correction to the significance level of 0.05 (i.e., three pairwise comparisons were performed per dependent variable). Note that only the pairwise comparisons of stair ambulation versus level walking were of interest; the pairwise comparison of stair ascent versus descent was beyond the scope of the present study.

3. Results

In the sagittal plane, stair ascent and descent both required greater peak moments at the knee and ankle joints in the first half of the stance phase, but a smaller peak moment at the ankle joint in the second half of stance when compared to level walking (Fig. 1). During stair ascent a hip extension moment was present throughout the stance phase. The peak hip extension and flexion moments were reduced during the first and second half of stance, respectively, in stair descent relative to level walking. In the frontal plane, a double-hump hip abduction moment was observed across all three functional tasks, but the magnitude of this moment was reduced during stair ascent.

The time histories of the predicted muscle forces were in general agreement with the recorded EMG linear envelopes for level walking and stair ambulation, except for SOL during stair descent and GMED during stair ascent (Fig. 2). Locomotor task had a