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Muscle function during gait is invariant to age when walking speed is controlled

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ABSTRACT

Older adults walk more slowly, take shorter steps, and spend more time with both legs on the ground compared to young adults. Although many studies have investigated the effects of aging on the kinematics and kinetics of gait, little is known about the corresponding changes in muscle function. The aim of this study was to describe and compare the actions of the lower-limb muscles in accelerating the body's center of mass (COM) in healthy young and older adults. Three-dimensional gait analysis and subject-specific musculoskeletal modeling were used to calculate lower-limb muscle forces and muscle contributions to COM accelerations when both groups walked at the same speed. The orientations of all body segments during walking, except that of the pelvis, were invariant to age when these quantities were expressed in a global reference frame. The older subjects tilted their pelves more anteriorly during the stance phase. The mean contributions of the gluteus maximus, gluteus medius, vasti, gastrocnemius and soleus to the vertical, fore-aft and mediolateral COM accelerations (support, progression and balance, respectively) were similar in the two groups. However, the gluteus medius contributed significantly less to support ($p < 0.05$) while the gluteus maximus and contralateral erector spinae contributed significantly more to balance ($p < 0.05$) during early stance in the older subjects. These results provide insight into the functional roles of the individual leg muscles during gait in older adults, and highlight the importance of the hip and back muscles in controlling mediolateral balance.

1. Introduction

Healthy older adults walk more slowly, take shorter steps, and spend more time with both legs on the ground (double-leg stance) than healthy young adults [1–5]. Peak hip extension angles are larger in older adults [1,5], whereas the moments and mechanical power generated by the ankle plantarflexor muscles are lower [2–4]. DeVita and Hortobagyi [6] noted that a reduction in the moment produced by the ankle plantarflexors coincided with an increase in the moment developed by the hip extensors, and suggested that lower-limb muscle work is re-distributed during gait in older adults.

However, differences in the gait characteristics of healthy young and older adults appear to diminish when walking speed is controlled [7]. Blanke and Hageman [8] and Silder et al. [9] found that step length was similar in healthy young and older adults when the two groups walked at a speed of 1.3 m/s. Jansen et al. [10] also reported no statistical age-related differences in step length, ground reaction forces, and a number of other gait characteristics when young and older subjects walked at a speed of 1.1 m/s.

Recent studies have identified the muscles that accelerate the body in the vertical, fore-aft and mediolateral directions (support, progression and balance, respectively [11]) for a range of walking speeds in healthy young adults [11–14]. The gluteus maximus, gluteus medius and vasti contribute significantly to support and slow forward progression during early stance, whereas the soleus and gastrocnemius support and propel the body in late stance. The gluteus medius also accelerates the body medially to counter the lateral accelerations produced by the vasti, soleus and gastrocnemius [11,12]. By comparison, very little is known about the functional roles of the lower-limb muscles during gait in older adults. Since most falls in older adults occur during walking [15], accurate knowledge of how individual muscles accelerate the center of mass of the body is important for the assessment of falls risk [16–18] and for the development of more effective exercise training programs to improve falls prevention in the elderly.

The goal of the present study was to investigate the effects of aging on lower-limb muscle function during walking. Three-dimensional gait analysis and subject-specific musculoskeletal modeling were used to determine muscle forces and muscle

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contributions to the center-of-mass (COM) accelerations when both groups walked at the same speed. The specific aim was to compare the contributions of individual lower-limb muscles to support, forward progression and mediolateral balance in healthy young and older adults. We hypothesized that there are no statistically significant age-related differences in lower-limb muscle function when walking speed is controlled.

2. Methods

Gait experiments were performed on 10 healthy young adults (age, 24.6 ± 2.5 yrs; height, 170 ± 8 cm; weight, 65.9 ± 7.9 kg; body mass index (BMI), 22.7 ± 1.6 kg/m^2) and 10 healthy older adults (age, 71.5 ± 6.3 yrs; height, 172 ± 9 cm; weight, 69.1 ± 13.5 kg; BMI, 23.3 ± 3.6 kg/m^2). Data were recorded in the Biomotion Laboratory at University of Melbourne after approval was obtained from the University's Human Research Ethics Committee. Each subject provided written informed consent prior to participating in the study. All subjects were able to walk unassisted and were free from any pathology that could influence their ability to walk normally. The older adults lived independently and maintained a minimum of five hours of physical activity per week.

Each subject walked at a prescribed speed of 1.4 m/s on level ground. This speed was chosen because it is close to the preferred walking speed for healthy young adults, and also because it represents a comfortable walking speed for healthy older adults [19]. Verbal feedback was provided after each trial to ensure that each subject's walking speed was as close as possible to 1.4 m/s. All older subjects walked comfortably at the prescribed speed. Kinematic, force-plate and muscle EMG data were recorded simultaneously for each walking trial. The three-dimensional positions of retro-reflective markers attached to each subject were recorded using a video motion capture system (VICON, Oxford Metrics Ltd., Oxford) with 9 cameras sampling at 120 Hz. A fourth-order Butterworth low-pass filter with a cut-off frequency of 4 Hz was used to smooth marker trajectories. Ground reaction forces were measured using three strain-gauged force plates (AMTI, Watertown, MA) mounted flush to the laboratory floor. Surface electrodes (Motion Laboratory Systems, Baton Rouge, LA) were attached to the right leg to record activity from six muscles: gluteus maximus, gluteus medius, vastus medialis, lateral hamstring, gastrocnemius and soleus. Placement of the EMG electrodes followed the procedures described in [20]. Ground-force and muscle-EMG data were sampled at 1080 Hz. Each subject performed five walking trials, and the trial for which walking speed was nearest to 1.4 m/s was selected for further analysis.

Scaled-generic musculoskeletal models were developed by scaling a 10-segment, 23 degree-of-freedom generic model of the body [21]. The head, arms and torso were represented as a single rigid segment, which articulated with the pelvis via a ball-and-socket joint. Each hip was modeled as a ball-and-socket joint, each knee as a hinge joint, and each ankle–subtalar joint complex as two intersecting hinge joints. The model was actuated by 54 muscle-tendon units, with each unit represented as a three-element Hill-type muscle in series with an elastic tendon [22]. Scaled-generic models were created by scaling segment lengths, segment inertial properties and muscle attachment sites in the generic model to individual subject anthropometry. The peak isometric force of each muscle was identical in the scaled and generic models, however, tendon slack length (l_s^T) and optimal muscle-fiber length (l_o^M) were scaled so that the ratio of these parameters remained the same for each muscle-tendon unit; that is, $[l_s^T/l_o^M]_{\text{generic}} = [l_s^T/l_o^M]_{\text{scaled}}$. The ratio l_s^T/l_o^M is a measure of muscle-tendon compliance in the model [22].

Inverse dynamics and static optimization were used to determine lower-limb muscle forces during gait. At each instant of the gait cycle, a set of optimal joint angles was computed by minimizing the sum of the squares of the differences between the positions of virtual markers defined in the musculoskeletal model and reflective markers attached to the subject. The calculated values of the joint angles and measured ground reaction forces were then applied to the model to compute the net moments exerted about the back, hip, knee and ankle joints. The net joint moments were decomposed into individual muscle forces by minimizing the sum of the squares of all muscle activations subject to the physiological bounds imposed by each muscle's force–length–velocity properties [23].

A pseudo-inverse force decomposition algorithm [24] was used to calculate the contributions of all muscle forces to the acceleration of the COM in the vertical, fore-aft, and mediolateral directions. The contribution of each muscle force to the three components of the ground reaction force was first found, and Newton's 2nd Law of Motion was then applied to determine each muscle's contribution to the COM accelerations. Data were averaged across all subjects for each group.

Independent *t*-tests were used to detect a statistically significant difference in each dependent variable (i.e., joint angles, net joint moments, and muscle contributions to COM accelerations) between the young and older groups. The results of the *t*-tests were further quantified using an '*h*-value'; $h = 1$ was assigned when the null hypothesis (i.e., no difference between the two groups) was rejected ($p < 0.05$), whereas $h = 0$ was assigned when $p > 0.05$. Data were analyzed for one complete gait cycle, but only the results for the stance phase are presented below.

3. Results

There were no statistically significant differences in mass, height, leg length, and BMI between the two groups ($p > 0.05$). Both groups exhibited similar gait characteristics with no significant differences in walking speed, step length, step frequency and step width ($p > 0.1$) (Table 1). However, the older subjects exhibited a longer duration for double-leg stance ($p = 0.04$).

Frontal- and transverse-plane joint angles were similar between the two groups, but the sagittal-plane joint angles differed significantly (Fig. 1A). The older subjects exhibited significant increases in back extension, anterior pelvic tilt, hip flexion and knee flexion for most of the stance phase, compared with the young subjects. However, the orientations of all the body segments, except that of the pelvis, were similar in the young and older subjects when these quantities were expressed in the ground reference frame. The sagittal-plane orientation of the pelvis was significantly different between the two groups, with the older subjects tilting their pelves more anteriorly throughout stance (Fig. 2A). There were no significant differences in the measured ground reaction forces between the two groups (Fig. 2B). The net joint moments were also similar for both groups, except for a significantly higher net hip extension moment exhibited by the older subjects during the first half of stance (Fig. 1B).

Overall, the timing of muscle forces predicted for each group was consistent with measured EMG activity (Fig. 3). Except for the gluteus maximus (GMAX) and hamstrings (HAMS), no significant differences were evident in the lower-limb muscle forces predicted for the two groups. The older subjects generated a higher GMAX

Table 1
Subject and trial characteristics. Data are presented as a group mean with standard deviations shown in parentheses. Only statistically significant differences between the two groups are shown ($p < 0.05$).

	Young group mean (SD)	Older group mean (SD)	*p*-Value ($p < 0.05$)
N	10	10	n.a.
Age (yrs)	24.62 (2.51)	71.45 (6.32)	0.0
Height (m)	1.70 (0.08)	1.72 (0.09)	–
Mass (kg)	65.93 (7.96)	69.11 (13.52)	–
BMI (kg/m^2)	22.7 (1.6)	23.3 (3.6)	–
Leg length (cm)	89.93 (5.36)	89.10 (4.67)	–
Trial speed (m/s^2)	1.40 (0.01)	1.42 (0.03)	–
Step length (cm)	73.20 (3.96)	72.20 (4.69)	–
Step frequency (steps/s)	1.90 (0.10)	1.98 (0.09)	–
Step width (cm)	8.53 (3.29)	6.78 (3.18)	–
Stance duration (% of gait cycle)	63.40 (1.62)	64.60 (0.68)	0.04
Swing duration (% of gait cycle)	36.60 (1.62)	35.37 (0.68)	0.04
Self-selected walking speed (m/s^2)	1.36 (0.20)	1.53 (0.15)	0.04