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Healthy older adults have insufficient hip range of motion and plantar flexor strength to walk like healthy young adults

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ABSTRACT

Limited plantar flexor strength and hip extension range of motion (ROM) in older adults are believed to underlie common age-related differences in gait. However, no studies of age-related differences in gait have quantified the percentage of strength and ROM used during gait. We examined peak hip angles, hip torques and plantar flexor torques, and corresponding estimates of functional capacity utilized (FCU), which we define as the percentage of available strength or joint ROM used, in 10 young and 10 older healthy adults walking under self-selected and controlled (slow and fast) conditions. Older adults walked with about 30% smaller hip extension angle, 28% larger hip flexion angle, 34% more hip extensor torque in the slow condition, and 12% less plantar flexor torque in the fast condition than young adults. Older adults had higher FCU than young adults for hip flexion angle (47% vs. 34%) and hip extensor torque (48% vs. 27%). FCUs for plantar flexor torque (both age groups) and hip extension angle (older adults in all conditions; young adults in self-selected gait) were not significantly $<100\%$, and were higher than for other measures examined. Older adults lacked sufficient hip extension ROM to walk with a hip extension angle as large as that of young adults. Similarly, in the fast gait condition older adults lacked the strength to match the plantar flexor torque produced by young adults. This supports the hypothesis that hip extension ROM and plantar flexor strength are limiting factors in gait and contribute to age-related differences in gait.

1. Introduction

Commonly reported age-related differences in gait may arise from physiological or neuromuscular limitations in older adults (McGibbon, 2003; Winter et al., 1990). Two particular impairments that may contribute to age-related differences in gait are reduced plantar flexor strength and reduced hip extension range of motion (ROM) (McGibbon, 2003). Many studies have reported reduced plantar flexor kinetics (e.g. peak torque, power generation and work) in older adults during gait (DeVita and Hortobagyi, 2000; Judge et al., 1996; Kerrigan et al., 1998; Monaco et al., 2009; Riley et al., 2001; Silder et al., 2008; Winter et al., 1990), which could be caused by reduced plantar flexor strength. Older adults may compensate for reduced plantar flexor kinetics with increased hip extensor power and work (DeVita and Hortobagyi, 2000; Monaco et al., 2009; Silder et al., 2008), or alternatively with increased hip flexor power and work in late stance (Cofre et al., 2011; Goldberg and Neptune, 2007; Judge et al., 1996; Monaco et al., 2009). In addition to kinetic differences, a number of studies have reported that older adults walk with a smaller peak hip extension angle than young adults (Kerrigan et al., 2001; Kerrigan et al., 1998; Lee et al., 2005; Monaco et al., 2009), possibly due to reduced hip extension ROM. Supporting this, a hip flexor stretching program that increased hip extension ROM in older adults also increased peak hip extension angle and step length during gait (Watt et al., 2011). Overall, previous studies suggest the hypothesis that older adults walk with lower plantar flexor peak torque and the hip more flexed than young adults because of limitations in plantar flexor strength and hip extension ROM, respectively. A few studies have examined the percentage of available strength used during level walking, calling it "muscular utilization ratio" (Nadeau et al., 1996, 1999; Requiao et al., 2005) or "functional demand" (Samuel et al., 2013). However, no previous studies have evaluated the percentage of strength or ROM used by healthy young and older adults, and whether these are in fact limiting factors that give rise to age-related differences in gait. Thus, this study examines the percentage of available strength and hip joint ROM, which we term functional capacity utilized (FCU), used by healthy young and older adults in level walking, in order to gain

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insight into the limitations underlying age-related differences in gait.

Most studies reporting differences in gait kinetics between young and older adults have also reported differences in speed and/or step length (DeVita and Hortobagyi, 2000; Judge et al., 1996; Kerrigan et al., 1998; Monaco et al., 2009; Riley et al., 2001; Winter et al., 1990). This may confound age differences in kinetics, as gait kinetics are affected by both speed (Graf et al., 2005; Kerrigan et al., 1998) and step length (Allet et al., 2011). Several studies have matched gait speed between age groups (Cofre et al., 2011; DeVita and Hortobagyi, 2000; Monaco et al., 2009), but older adults may still walk with shorter step length and higher cadence than young adults. When speed and step length do not differ between age groups, age differences remain in gait kinetics (Silder et al., 2008), indicating that age differences arise from factors besides older adults choosing a different speed and step length. However, to understand how functional impairments affect age differences in gait, it may be necessary to control both speed and step length to prevent confounding by common age differences in spatio-temporal characteristics. Thus to account for the effects of both speed and step length, this study examines controlled gait with fixed speeds and step length in addition to self-selected gait.

2. Methods

2.1. Participants

Ten young (ages 20–31) and 10 older (ages 75–86) healthy adult volunteers participated in functional testing and gait testing (Table 1). Participants could walk independently and reported no musculoskeletal, neurological, cardiovascular, or cognitive disorders that might affect gait. The study was approved by the Virginia Tech Institutional Review Board, and all participants provided written informed consent.

2.2. Measurement

During functional testing, maximum isometric joint torques in hip extension, hip flexion, and plantar flexion, as well as hip extension and hip flexion ROM were measured on the right lower extremity using a Biodex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, New York, USA). Plantar flexion testing was performed with the standard manufacturer attachment, while hip testing was performed in an upright position with the body stabilized in a custom-built frame and the knee immobilized at about 0° of knee flexion. Maximum isometric joint torque was measured as the largest of three or more maximum voluntary exertions performed at the approximate joint angles at which maximum isometric joint torques occur (Anderson et al., 2007). If a participant's joint ROM was less than the target angle, strength was tested at the limit of the ROM. Specifically, hip extension torque was tested at a target angle of 68° of hip flexion (ROM limited to 56°, 60°, 60°, 60°, 62°, 62° and 65° in seven participants), hip flexion torque was tested at a target angle of 15° of hip extension (ROM limited to 11° and 13° in two participants), and plantar flexion torque was tested at a target angle of 26° of dorsiflexion (ROM limited to 23° in one participant). Joint torque and angle were recorded at 200 Hz and low pass filtered at 5 Hz (Anderson et al., 2007) to attenuate electromechanical noise in the data. Prior to strength testing, each joint was passively moved through two full cycles of joint motion at 5°/s while participants remained relaxed, and this data was used to correct strength measurements for gravitational moments and model passive elastic joint torques (Anderson et al., 2010). Hip ROMs were determined by manually moving the hip joint to the participant's limits of motion.

Each participant performed gait trials on an 8 m long walkway under self-selected and controlled (slow and fast) conditions. Participants practiced each condition until comfortable with the task prior to recording data. First, a single trial of self-selected gait was recorded, followed by slow and fast conditions presented in random order. Target speeds (slow: 1.1 m/s; fast: 1.5 m/s), and step length (0.65 m for both speeds) were representative of values reported in the literature for self-selected gait in young and older adults (Judge et al., 1996; Kerrigan et al., 1998; Silder et al., 2008; Winter et al., 1990). A moving belt beside the walkway and stripes painted across walkway provided cues of target speed and step length, respectively. Trials were repeated if estimated speed was not within 5% of the target speed. Ground reaction forces were sampled at 1000 Hz from a six degree-of-freedom force platform (Advanced Mechanical Technology Inc., Watertown, MA, USA) in the center of the walkway. Kinematics were determined from four reflective markers on each foot, five on each shank, five on each thigh, four on the pelvis, and four on the upper body sampled at 100 Hz using a six-camera VICON 460 motion analysis system (VICON Motion Systems Inc., Lake Forest, CA, USA).

Table 1
Mean (SD) subject characteristics and functional measures by age group. Maximum isometric joint torques are normalized by body mass. Hip extension and flexion ROMs were measured relative to the anatomical position.

	Young ($N=10$)	Older ($N=10$)	% Difference
Age (years)	23.9 (3.3)	80.3 (4.0)	
Sex (M/F)	5/5	5/5	
Body mass (kg)	61.7 (7.3)	65.2 (10.5)	6
Height (m)	1.65 (0.09)	1.63 (0.08)	0
Hip ROM (deg):			
Extension	28.0 (7.8)	19.7 (5.2)	−30*
Flexion	73.5 (8.4)	69.1 (10.9)	−6
Total	101.5 (13.9)	87.1 (15.2)	−14*
Maximum isometric joint torque (N-m/kg):			
Hip Extension	4.51 (0.94)	3.02 (0.85)	−33*
Hip Flexion	2.67 (0.51)	1.76 (0.23)	−34*
Plantar Flexion	2.64 (0.59)	2.12 (0.25)	−20*

* Significant age difference ($p<0.05$).

2.3. Data processing

The model of Delp et al. (1990) was adapted to create an individualized, eight segment, 27 degree-of-freedom, model of the upper body, pelvis and lower extremities for each participant in OpenSim (Delp et al., 2007). Anthropometry measurements were used to estimate segment masses, center of mass locations and mass moments of inertia (de Leva, 1996; Pavol et al., 2002). Joint center positions were determined by functional methods (Piazza et al., 2004) and data from a calibration trial in which each participant moved each joint through its degrees of freedom.

Speed and step length were evaluated from marker data, and one trial best representing the desired speed and step length was selected for each controlled gait condition. Ground reaction forces and marker motion data were low pass filtered at 7 Hz and center of pressure position was determined (Winter, 2005). Inverse kinematics and 3D inverse dynamics were performed in OpenSim to estimate internal joint torques over one full swing and stance cycle of the right lower extremity.

Primary variables of interest during gait were peak hip extension angle (HEA), peak hip flexion angle (HFA), total hip angular excursion (THA), peak hip extension torque (HET), peak hip flexion torque (HFT) and peak plantar flexion torque (PFT) (Fig. 1). For hip angles, FCU was calculated as

$$\mathrm{FCU}=\frac{A}{R}\times 100 \tag{1}$$

where A is the hip angle of interest during gait (HEA, HFA or THA), and R is the corresponding measured ROM (hip extension ROM, hip flexion ROM, or total hip ROM). For joint torques, FCU was calculated as

$$\mathrm{FCU}=\frac{T}{S+P}\times 100 \tag{2}$$

where T is the peak joint torque of interest during gait (HET, HFT or PFT), and S and P are the corresponding active and passive components of available joint strength, respectively. S was determined by adjusting measured isometric torque for joint angle and angular velocity at the time of peak joint torque using a model relating joint torque, angle and angular velocity (Anderson et al., 2007). Specifically:

$$S=\begin{cases} S_{ISOMETRIC}\ \cos\left(C_2(\theta-C_3)\right)\left(\frac{2C_4C_5+\dot{\theta}(C_5-3C_4)}{2C_4C_5+\dot{\theta}(2C_5-4C_4)}\right) & \dot{\theta}\geq 0 \\ S_{ISOMETRIC}\ \cos\left(C_2(\theta-C_3)\right)\left(\frac{2C_4C_5-\dot{\theta}(C_5-3C_4)}{2C_4C_5-\dot{\theta}(2C_5-4C_4)}\right)(1-C_6\dot{\theta}) & \dot{\theta}<0 \end{cases} \tag{3}$$

where $S_{ISOMETRIC}$ is measured maximum isometric joint torque, $\theta$is joint angle at the time of peak torque, and $\dot{\theta}$ is joint angular velocity at the time of peak torque. Values of the coefficients C_2–C_6 were taken from the literature based on participant age and sex (Anderson et al., 2007). Passive joint torque P was determined using the model:

$$P=B_1e^{k_1\theta}+B_2e^{k_2\theta} \tag{4}$$

with coefficients B and k determined from passive motion data as previously described (Anderson et al., 2010). Measurement errors were found in hip ROM data for one participant and plantar flexor strength for two participants, and these values were imputed based on age and sex when calculating FCUs.