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Advanced age affects the individual leg mechanics of level, uphill, and downhill walking

Jason R. Franz *, Rodger Kram

Department of Integrative Physiology, University of Colorado, Boulder, CO, USA

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ABSTRACT

Advanced age brings biomechanical changes that may limit the uphill and/or downhill walking ability of old adults. Here, we investigated how advanced age alters individual leg mechanics during level, uphill, and downhill walking. We hypothesized that, compared to young adults, old adults would exhibit: (1) reduced trailing leg propulsive ground reaction forces (GRFs) and positive work rates during uphill walking, and (2) reduced leading leg braking ground reaction forces and negative work rates during downhill walking. We calculated the individual leg mechanical work performed by 10 old (mean $\pm$ SD, age: 72 ± 5 yrs) and 11 young (age: 26 ± 5 yrs) adults walking at 1.25 m/s on a dual-belt force-measuring treadmill at seven grades (0° and $\pm3^{\circ}$, $\pm6^{\circ}$, $\pm9^{\circ}$). As hypothesized, old adults exhibited significantly reduced propulsive GRFs (e.g., -21% at $+9^{\circ}$) and average trailing leg positive work rates (e.g., -26% at $+9^{\circ}$) compared to young adults during both level and uphill walking. Old adults compensated by performing greater positive work than young adults during the subsequent single support phase. In contrast, we reject our second hypothesis. We found no differences in braking GRFs or negative work rates between old and young adults. However, old adults exhibited significantly reduced second peak perpendicular GRFs during downhill walking compared to young adults. Our findings most notably identify how advanced age may impair uphill walking ability and thus independence and quality of life.

1. Introduction

Advanced age (i.e., 65+ years) brings biomechanical changes that negatively affect walking ability, independence, and quality of life. An abundant literature describes how these age-related changes manifest during level walking. For example, old adults exhibit a distal to proximal redistribution of mechanical power production from the muscles of the ankle to the muscles of the hip (DeVita and Hortobagyi, 2000; Savelburg et al., 2007; Cofre et al., 2011). This greater reliance on hip muscles coincides with reduced trailing leg propulsion during double support compared to young adults walking at the same speed (Ortega and Farley, 2007; Hernandez et al., 2009). Another pervasive consequence of aging is sarcopenia, the loss of skeletal muscle mass, which is often associated with functional declines in strength (e.g., Baumgartner et al. 1998). Together, hip muscle reliance, reduced trailing leg propulsion, and leg muscle weakness may limit the uphill and/or downhill walking ability of old adults. However, to the best of our knowledge, there are no published studies that compare the biomechanics of uphill and downhill walking in old vs. young adults.

The biomechanics of individual leg function during uphill and downhill walking in young adults are strikingly different from walking over level ground. During level walking, the leading and trailing legs simultaneously and nearly exclusively perform negative and positive mechanical work during double support, respectively (Donelan et al., 2002). However, we recently discovered that both the leading and trailing legs of young adults perform progressively greater positive work with steeper uphill grade and greater negative work with steeper downhill grade (Franz et al., 2012). Thus, in contrast to level walking, up to one-third of the mechanical work performed during the double support phase of uphill (positive work) or downhill (negative work) walking is performed by the leading or trailing leg, respectively. Electromyographic (EMG) data from young adults provides additional insight into the coordinated performance of leading and trailing leg mechanical work during uphill and downhill walking (Lay et al., 2007; Franz and Kram, 2012). These data suggest that during uphill walking, trailing leg ankle extensor muscles are aided by leading leg hip and knee extensor muscles to perform the greater positive work to raise the center of mass (CoM). Further, during downhill walking, the leading and trailing leg knee extensor muscles largely perform the greater negative work to lower the CoM.

* Correspondence to: Department of Integrative Physiology, UCB 354, University of Colorado, Boulder, CO 80309-0354, USA. Tel.: +1 303 492 7984, +1 410 303 6883; fax: +1 303 492 4009.
E-mail address: jason.franz@colorado.edu (J.R. Franz).

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Age-related biomechanical changes known to manifest during level walking may adversely affect individual leg mechanics during uphill and/or downhill walking. The performance of trailing leg positive work is particularly important to meet the propulsive demands of uphill walking and raise the CoM with each step. Reduced trailing leg propulsion and leg muscle weakness in old adults could compromise their uphill walking ability. Moreover, as a consequence of their disproportionate recruitment of hip muscles, old adults approach the maximum voluntary isometric capacity of their gluteus maximus (a hip extensor muscle) during uphill walking (Franz and Kram, in press). Thus, unlike in young adults walking uphill, the hip muscles of old adults may be unable to adequately assist the trailing leg ankle extensor muscles and perform the positive work necessary to raise the COM. During downhill walking, knee extensor (quadriceps) muscle atrophy and weakness commonly observed with advanced age (Trappe et al., 2001) could compromise the ability of the leading leg of old adults to perform the considerable negative work necessary to eccentrically lower the CoM with each step.

Here, we investigated how advanced age affects individual leg mechanics during level, uphill, and downhill walking. We hypothesized that, compared to young adults, old adults would exhibit: (1) reduced trailing leg propulsive ground reaction forces (GRFs) and positive work rates during uphill walking, and (2) reduced leading leg braking ground reaction forces and negative work rates during downhill walking.

2. Methods

2.1. Subjects

10 old adults (6F/4M, mean ± SD, age: 72 ± 5 yrs, height: 1.70 ± 0.10 m, mass: 65.0 ± 13.3 kg) and 11 young adults (5F/6M, age: 26 ± 5 yrs, height: 1.76 ± 0.10 m, mass: 71.0 ± 12.3 kg) participated. All subjects were healthy and exercised regularly. Prior to participating, subjects completed a health questionnaire based upon recommendations of the American College of Sports Medicine (2006). We excluded subjects for any of the following criteria: BMI ≥ 30, sedentary lifestyle, first degree family history of coronary artery disease, cigarette smoking, high blood pressure, high cholesterol, diabetes or prediabetes, orthopedic or neurological condition, or taking medication that causes dizziness. All subjects scored at the highest possible mobility on the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994). All subjects gave written informed consent as per the University of Colorado Institutional Review Board.

2.2. Experimental procedures

Subjects began each of four experimental sessions by walking at 1.25 m/s on a level, motorized treadmill (model 18-60, Quinton Instruments, Seattle, WA, USA) for 5 min. Subjects then walked at 1.25 m/s on a dual-belt, force-measuring treadmill (Kram et al., 1998) for 2 min on the level (session 1), and both uphill and downhill (sessions 2–4) at one of three grades (3°, 6°, 9°; i.e., 5.2%, 10.5%, 15.7%). A force platform (model ZBP-7124-6-4000, Advanced Mechanical Technology, Inc., Watertown, MA, USA) was mounted under one side of the treadmill (Fig. 1). For uphill and downhill walking trials, we mounted both sides of the treadmill in parallel on custom-made aluminum wedges as described in earlier studies (Gottschall and Kram, 2006; Franz et al., 2012). Subjects completed experimental sessions on four separate days and in the same order of conditions due to the lengthy process of interchanging the aluminum wedges. We reversed the treadmill belt direction so that subjects could walk both uphill and downhill at one grade during a single session. We recorded the ground reaction force (GRF) components perpendicular, parallel, and lateral to the treadmill surface produced by one leg at 1000 Hz during the last 30 s of each trial. Specifically, we recorded right leg GRFs during level and uphill walking and left leg GRFs during downhill walking.

2.3. Data analysis

A custom script written in MATLAB (Mathworks, Inc., Natick, MA, USA) processed all data. We digitally filtered the GRF signals using a recursive fourth-order Butterworth filter with a low-pass cutoff frequency of 20 Hz. A 10% body-weight threshold on the perpendicular force signal identified the stance phase GRFs, which we then averaged over 15 consecutive strides per condition. Phase-shifting the resulting GRF profiles by 50% emulated the forces produced by the contralateral leg (symmetry assumed, see Seeley et al. (2008, Hernandez et al. (2009), Burnett et al. (2011)).

We used the individual limbs method (ILM) to calculate the mechanical work rates for each leg from the stride-averaged GRF data (Donelan et al., 2002). We tailored the ILM to account for the treadmill grade for uphill and downhill walking trials, as described in detail previously (Franz et al., 2012). Briefly, we calculated each leg's instantaneous mechanical work rate as the dot product of the three-dimensional GRF and center of mass (CoM) velocity (determined by integrating the GRF signals with respect to time). By then restricting the cumulative time-integrals of the instantaneous mechanical work rate curves to the intervals over which the integrand was positive or negative, we determined the positive and negative mechanical work performed by each leg, respectively. We performed these integrations over the following time intervals of interest: double support, single support, and a complete step (heel-strike of one leg to heel-strike of the contralateral leg). We calculated the average mechanical work rate (reported in W/kg) by dividing each measure of individual leg mechanical work (J/kg) by the average step duration (s).

Finally, we prepared CoM hodographs, plots of sagittal plane CoM velocity components, over an average step for level, uphill, and downhill walking. Adamczyk and Kuo (2009) first showed that CoM hodographs can elucidate how individual leg work redirects and restores the CoM velocity to upward and forward during the transition from one period of single support to the next. Thus, we used CoM hodographs to graphically demonstrate how individual leg work differentially affects the CoM velocity redirection in old vs. young adults during level, uphill, and downhill walking. To describe the CoM hodographs, we use the terms pre- and post-transition, defined by other authors (Adamczyk and Kuo, 2009; Soo and Donelan, 2012) as the time at which the perpendicular velocity reaches a minimum (just prior to heel-strike) and maximum (just following toe-off) during a step, respectively.

We compared step and double support durations, and the average mechanical work rates in old adults to those obtained by re-analyzing young adult GRF data from our study published previously (Franz et al., 2012). A repeated measures analysis of variance tested for significant main effects of age and age by grade interactions with a $p < 0.05$ criterion. When a significant main effect of age was found, we performed independent samples t-tests to determine at which grade(s) the differences occurred.

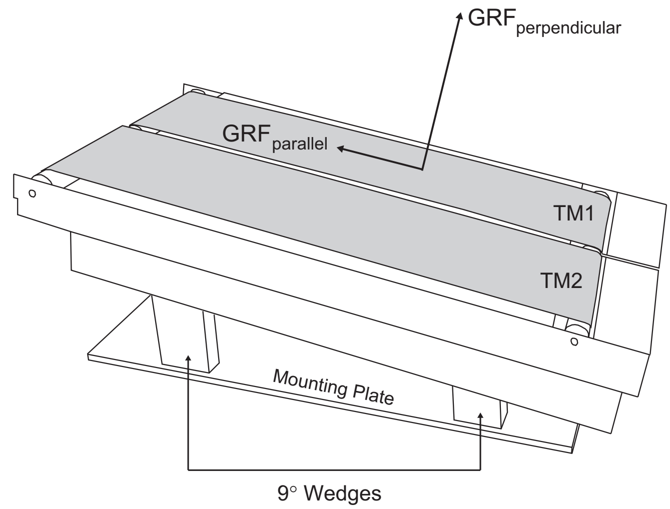

Fig. 1. Dual-belt force-measuring treadmill mounted at 9°. A force platform mounted under treadmill TM1 recorded the perpendicular, parallel, and lateral (not shown) components of the ground reaction force (GRF) produced by a single leg. Assuming symmetry, we phase shifted those GRFs to emulate the forces produced by the contralateral leg. TM: treadmill.

3. Results

Compared to young adults (Y) walking at the same speed, old adults (O) walked with significantly smaller peak parallel propulsive GRFs (e.g., O: 24.3 ± 3.2%BW vs. Y: 30.8 ± 1.5%BW at +9°) and second peak perpendicular GRFs (e.g., O: 94.2 ± 12.7%BW vs. Y: 112.1 ± 6.7%BW at +9°) during level and uphill conditions (Fig. 2). A significant age by grade interaction revealed that old adults,