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

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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 $\pm 3^\circ$, $\pm 6^\circ$, $\pm 9^\circ$). 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.

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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.

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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

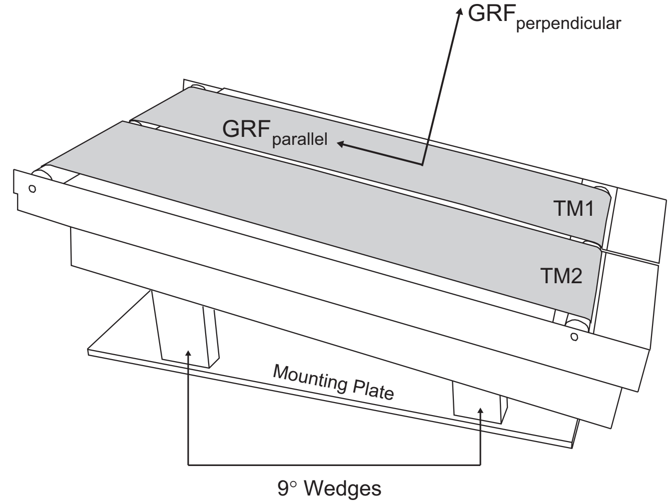

Fig. 1. Dual-belt force-measuring treadmill mounted at 9°. A force platform mounted under treadmill TM1 recorded the perpendicular, parallel, and lateral (not shown) components of the ground reaction force (GRF) produced by a single leg. Assuming symmetry, we phase shifted those GRFs to emulate the forces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

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.

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,

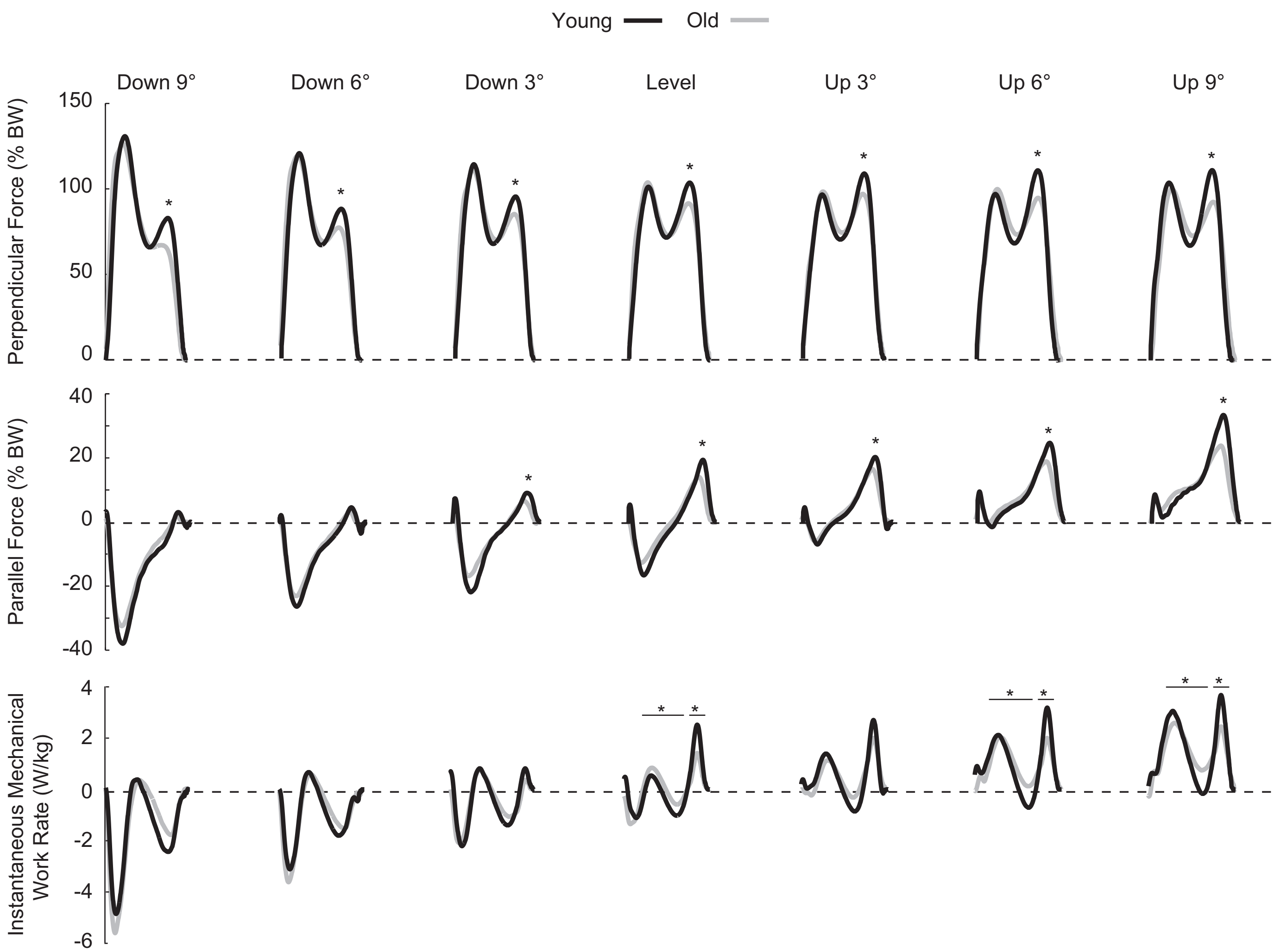

Fig. 2. Mean perpendicular and parallel ground reaction forces (GRFs) and individual leg instantaneous mechanical work rates for old and young adults normalized to body mass. All curves represent one stance phase normalized to the gait cycle. Asterisks (*) indicate a significant difference between old and young adults ($p<0.05$). For instantaneous work rates, asterisks correspond to regions of the stance phase (i.e., single support, double support) over which the average work rates differed significantly in old vs. young adults. We identified these regions from the individual leg GRFs.

unlike young adults, did not increase their second peak perpendicular GRF at uphill grades steeper than 3° ($p<0.01$). During downhill walking, old adults exhibited significantly smaller second peak perpendicular GRFs than young adults (e.g., 70.8 ± 10.0%BW vs. 84.6 ± 5.4%BW at $-9°$, $p<0.01$).

Old adults also exhibited significantly reduced trailing leg positive work rates during the double support phase of level and uphill walking (Fig. 2, Fig. 3A, Table 1). Old adults compensated with greater positive work rates than young adults during the subsequent single support phase (Fig. 2, Fig. 3B, Table 1). Consequently, average positive work rates over a complete step were largely unaffected by age (Fig. 3C). Importantly, we found no differences in step and double support durations that could have contributed to the age differences in GRFs and mechanical work rates during uphill walking (step duration, $p=0.07$; double support duration, $p=0.28$) (Table 2). Further, we found no significant effect of age on negative work rates during downhill walking.

Fig. 4 graphically shows how the individual legs of old and young adults served to restore and redirect the CoM velocity upward and forward during the transition from one period of single support to the next during level, uphill, and downhill walking. The most prominent age difference in these CoM hodographs was that old adults exhibited smaller CoM velocity fluctuations than young adults. That is, old adults' CoM velocity deviated considerably less from the mean walking speed. These more conserved patterns in old adults became progressively more evident with steeper uphill grade. Further, consistent with reduced trailing leg propulsive function, old adults did not demonstrate the progressively greater pre- and post-transition CoM velocities at steeper uphill grades exhibited by young adults.

4. Discussion

Our study provides the first biomechanical insights into how advanced age affects the individual leg mechanics of uphill and downhill walking. We accept our first hypothesis that trailing leg propulsive function in old adults would be compromised during uphill walking. At the same walking speed, old adults exhibited significantly reduced double support trailing leg positive work rates and propulsive GRFs compared to young adults. Old adults compensated by performing greater positive work than young adults during single support. However, in contrast to our second hypothesis, we found no age differences in braking GRFs or negative work rates during downhill walking.

Ankle extensor muscles are largely responsible for performing trailing leg positive work (Zelik and Kuo, 2010). Consequently, reduced trailing leg propulsive function in old adults results from

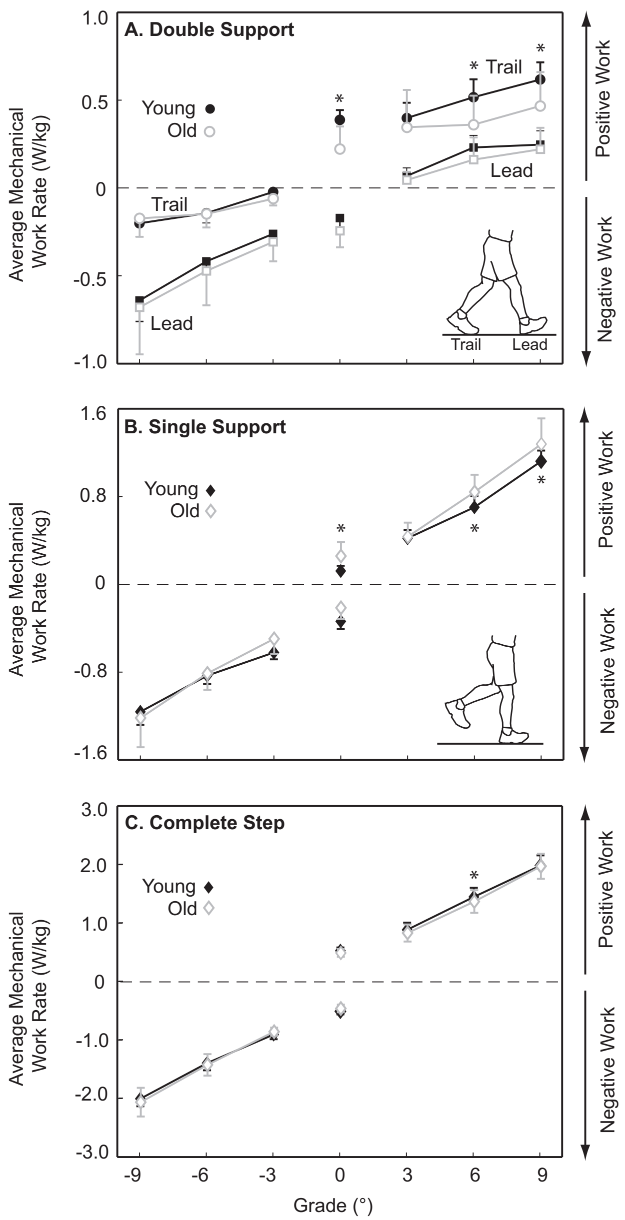

Fig. 3. Mean (standard deviation) individual leg positive (uphill) and negative (downhill) mechanical work rates during (A) double support, (B) single support, and (C) over a complete step normalized to body mass. Asterisks (*) indicate a significant difference between old and young adults ($p < 0.05$).

either distal muscle weakness secondary to sarcopenia and/or the disproportionate recruitment of hip vs. ankle muscles (Christ et al., 1992; DeVita and Hortobagyi, 2000; Savelburg et al., 2007; Cofre et al., 2011). Silder et al. (2008) reported that ankle muscle weakness is correlated with reduced ankle extensor power generation during walking in old adults. However, we recently showed that a comparable group of old adults retained the ability to double their ankle extensor muscle activity during uphill vs. level walking despite their having 30–37% less isometric leg strength compared to young adults (Franz and Kram, in press). We infer from those findings that old adults have an underutilized reserve for recruiting ankle muscles and further that ankle muscle weakness contributes only marginally to the propulsive impairments exhibited by old adults during level walking. We suspect that neural adaptations with age bring the disproportionate recruitment of proximal vs. distal leg muscles in old adults (Franz and Kram, in press) which preclude their performing as much trailing leg positive work as young adults. During both level and uphill walking, our observation that old adults compensate for reduced trailing leg positive work during the single support phase is consistent with a greater recruitment of more proximal leg muscles in old vs. young adults. However, the effect of leg muscle weakness on reduced trailing leg positive work in old adults may be exacerbated on steeper uphill grades.

Because knee extension weakness may impair the leading leg's ability to eccentrically control lowering the CoM (Ikezoe et al., 2011), we were surprised to find that old and young adults demonstrated equivalent braking GRFs and negative work rates during downhill walking. However, old adults performed the same amount of negative work during downhill walking differently (i.e., with smaller second peak perpendicular GRFs compared to young adults). Because we found no differences in step kinematics, it follows that the reduced GRFs exhibited by old vs. young adults are offset by greater negative CoM velocities during downhill walking (Fig. 4). Although we reject our second hypothesis, our findings do not exclude an effect of muscle weakness on the downhill walking ability of old adults. Indeed, such an impairment may underlie the reduced perpendicular GRF during downhill walking, which implies that old adults are not decelerating the substantial downward motion of the CoM during the late stance phase as much as young adults.

As a complement to measures of mechanical work, we graphically demonstrated how uphill and downhill walking differentially affect the CoM velocity redirection in old vs. young adults (Fig. 4). For both old and young adults, uphill and downhill grades skew the CoM hodographs in directions that correspond to performing greater positive and negative work, respectively. However, old adults exhibited more conserved patterns of CoM

Table 1
Mean (standard deviation) values of mechanical work rates (W/kg).

		Negative work rate				Positive work rate			
		−9°	−6°	−3°	0°	0°	3°	6°	9°
DS_T	O	−0.17 ± 0.09	−0.15 ± 0.08	−0.06 ± 0.04	−0.01 ± 0.01	**0.21 ± 0.12**	0.33 ± 0.17	**0.35 ± 0.13**	**0.46 ± 0.15**
	Y	−0.20 ± 0.08	−0.14 ± 0.05	−0.02 ± 0.02	0.00 ± 0.00	**0.39 ± 0.05**	0.40 ± 0.07	**0.51 ± 0.08**	**0.62 ± 0.06**
DS_L	O	−0.66 ± 0.21	−0.46 ± 0.15	−0.31 ± 0.12	−0.25 ± 0.11	0.00 ± 0.01	0.04 ± 0.05	0.15 ± 0.11	0.22 ± 0.11
	Y	−0.64 ± 0.11	−0.42 ± 0.11	−0.26 ± 0.05	−0.17 ± 0.06	0.02 ± 0.01	0.07 ± 0.04	0.23 ± 0.06	0.25 ± 0.07
SS	O	−1.25 ± 0.38	−0.83 ± 0.21	−0.50 ± 0.13	−0.21 ± 0.09	**0.27 ± 0.16**	0.45 ± 0.17	**0.86 ± 0.23**	**1.29 ± 0.25**
	Y	−1.16 ± 0.11	−0.83 ± 0.05	−0.62 ± 0.06	−0.34 ± 0.06	**0.12 ± 0.05**	0.42 ± 0.07	**0.70 ± 0.10**	**1.12 ± 0.09**

DS_T: double support, trailing leg. DS_L: double support, leading leg. SS: single support. O: old. Y: young. Bolded values indicate a significant difference between old and young adults ($p < 0.05$).

Table 2
Mean (standard deviation) values of step and double support times (s).

		−9°	−6°	−3°	0°	3°	6°	9°
t_{step}	Young	0.50 (0.03)	0.52 (0.03)	0.54 (0.03)	0.54 (0.03)	0.55 (0.04)	0.54 (0.04)	0.54 (0.04)
	Old	0.47 (0.06)	0.49 (0.06)	0.49 (0.05)	0.51 (0.05)	0.50 (0.06)	0.50 (0.07)	0.48 (0.06)
t_{ds}	Young	0.10 (0.01)	0.10 (0.01)	0.10 (0.01)	0.12 (0.01)	0.12 (0.01)	0.12 (0.01)	0.12 (0.01)
	Old	0.08 (0.03)	0.09 (0.02)	0.10 (0.02)	0.11 (0.02)	0.11 (0.02)	0.11 (0.03)	0.12 (0.03)

t_{step}: step time. t_{ds}: double support time. We did not find any significant effect of age on step and double support times.

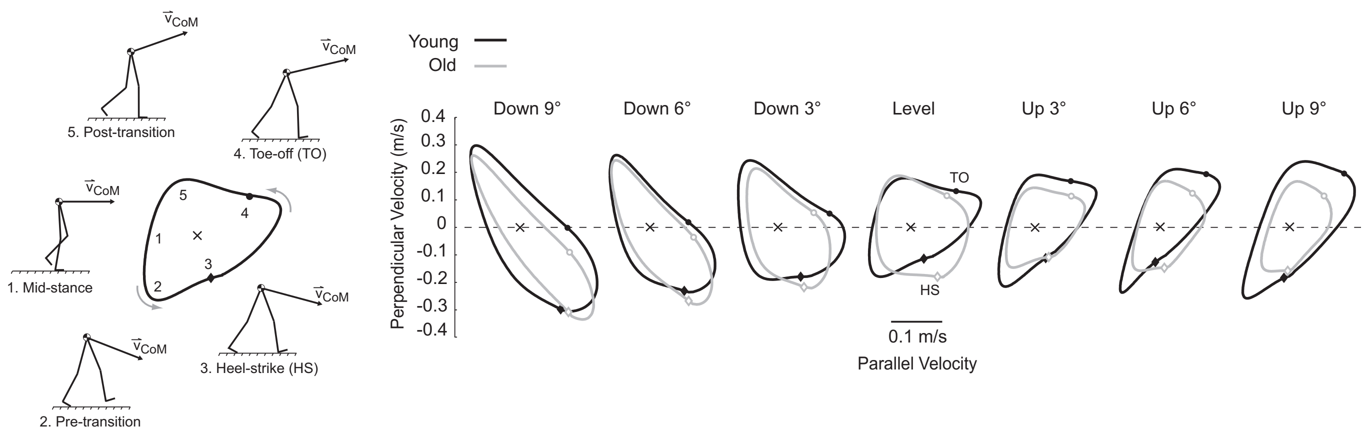

Fig. 4. Plots of the sagittal plane center of mass (CoM) velocity components (CoM hodographs) over an average step during level, uphill, and downhill walking. Mean walking speed (perpendicular: 0 m/s, parallel: 1.25 m/s) is indicated by ×. For level walking, the CoM velocity is directed downward and forward at the end of single support. The mechanical work performed by the individual legs serves to redirect and restore the CoM velocity to upward and forward during the transition from one period of single support to the next. For level, uphill, and downhill walking, the redirection began as the perpendicular CoM velocity reached a minimum and was complete when the perpendicular CoM velocity had reached a maximum. For our purposes, CoM hodographs visually demonstrate how individual leg work affected the CoM velocity redirection differently in old vs. young adults during level, uphill, and downhill walking.

velocity fluctuations than young adults. This may at first be interpreted as advantageous, reflecting a smoother walking pattern than young adults particularly during uphill walking. However, we propose that the observed age differences signify a more restricted pattern of CoM velocity change constrained by the reduced propulsive mechanics of old adults. Moreover, imposing a smoother pattern of CoM velocity over a step in young adults can impair the exchange of mechanical energy and double the metabolic cost of level walking (Ortega and Farley, 2005). This suggests that old adults adopt a pattern of CoM motion during uphill walking which may incur a metabolic penalty compared to young adults.

Our old adult subjects exhibited what may be considered adverse biomechanical changes despite their exceptional physical activity and fitness. In this context, and based on our observations, it is reasonable to presume that the old adults in this study were not aerobically limited during any of the walking trials. In contrast, due to the age-related decline in aerobic capacity (Fitzgerald et al., 1997), more sedentary old adults can approach what has been termed "aerobic frailty" ($\dot{V}O_{2max} \leq 18$ ml/kg/min) (Booth and Zwetsloot, 2010), which makes uphill walking exhausting. In contrast, more active old adults who slow the decline in aerobic capacity may exhibit "biomechanical locomotor frailty" prior to their experiencing aerobic limitations. To characterize biomechanical locomotor frailty, future research could identify the uphill grade and/or walking speed at which the greater positive work performed by old adults during single support fails to compensate for trailing leg propulsive deficits.

One possible limitation of this study is that we measured only the forces produced by a single leg. However, other authors have confirmed the bilateral symmetry of GRF measures during walking in young adults (Seeley et al., 2008; Burnett et al., 2011). We are not aware of any published studies that suggest that healthy old adults walk more or less symmetrically than young adults. In addition, the total mechanical work performed during walking includes internal work to accelerate and decelerate the limbs with respect to the CoM (Cavagna and Kaneko, 1977) that is not quantified by the ILM. However, the internal work of the legs changes relatively little with uphill or downhill grade (Minetti et al., 1993) and has been found to be similar in old vs. young adults during level walking (Mian et al., 2006). Finally, our findings are also limited to active and healthy old adults. We would expect even greater deficits in the propulsive mechanics of sedentary and/or frail old adults due to the more direct effects of advanced sarcopenia and leg muscle disuse and weakness.

This study reveals how advanced age affects individual leg mechanics during level, uphill, and downhill walking. We find that even active and healthy old adults exhibit compromised trailing leg mechanical power generation during level and uphill walking. We also find that old adults exhibit significantly reduced second peak perpendicular GRFs during downhill walking compared to young adults which we suspect may result from knee extensor (quadriceps) muscle weakness. Our findings most notably identify how advanced age may impair uphill walking ability and point to specific targets (e.g., trailing leg propulsive forces during uphill walking) for biomechanical interventions that might prolong independent mobility in old adults.

Conflict of interest statement

The authors have no conflicts of interest to disclose.

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