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ScienceDirect
Clinical Biomechanics 22 (2007) 574–580
CLINICAL BIOMECHANICS
www.elsevier.com/locate/clinbiomech

The effect of step length on young and elderly women's ability to recover balance

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Received 6 May 2006; accepted 26 January 2007

Abstract

Background. Stepping is a common technique for balance recovery. Previous studies have shown that elderly adults are less able than young adults to recover balance by stepping, apparently due to reductions in step length and step speed. We sought to clarify these relations by testing the hypothesis that step length and step speed affect the ability of young and elderly women to recover balance.

Methods. During experimental trials, we measured the maximum release angle where participants could recover balance using a single forward step of length 15%, 25%, or 35% body height.

Findings. Both step length and age associated with recovery ability ($P < 0.001$). When step length increased from 15% to 25% body height, the maximum release angle increased by 36% in young participants and 31% in elderly. When step length increased from 25% to 35% body height, the maximum release angle increased 23% in young and 6% in elderly. At all step lengths, maximum release angles were greater in young than elderly women (by 21% at 15% body height, 30% at 25% body height, and 51% at 35% body height). For all but the 15% body height condition, recovery ability correlated with step contact time, which averaged 50–100 ms faster in young than in elderly.

Interpretation. The ability of young and elderly women to recover balance by stepping is enhanced by taking larger and quicker steps. This should be considered in balance assessment and training.

Keywords: Posture; Motor control; Perturbation; Falls; Hip fracture; Biomechanics; Aging

1. Introduction

In the event of loss of balance, humans commonly recover a stable upright stance by taking a step (Maki et al., 2003). Furthermore, when the strength of the destabilizing perturbation is increased, participants respond by taking quicker and larger steps (Do et al., 1982; Hsiao and Robinovitch, 1999; Luchies et al., 1994; Thelen et al., 1997; Wojcik et al., 1999). Such adaptations make sense from a biomechanical perspective, since larger perturbations will tend to thrust the body's center-of-gravity away from the base-of-support more rapidly, and balance recovery will occur only if the stepping foot (and the border of the base of support) is quickly relocated ahead of the center-of-gravity (Maki and Mcilroy, 1999; Pai and Patton, 1997). Furthermore, larger steps generate larger force impulses between the foot and the ground at landing (King et al., 2005).

Studies have also shown that elderly adults are less able than young adults to recover balance by stepping, apparently due to reductions in both step length and step speed (Hageman and Blanke, 1986; Medell and Alexander, 2000; Oberg et al., 1993). However, we currently have an incomplete understanding of the effect of step length and step speed on ability to recover balance by stepping. This is because previous studies on balance recovery by stepping either placed no restrictions on step characteristics (Thelen

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doi:10.1016/j.clinbiomech.2007.01.013

et al., 1997; Wojcik et al., 1999), or because participants were not challenged to the limits of their ability (Do et al., 1982; Maki and Mcilroy, 1999; King et al., 2005). Presumably, one should be able to recover from a specific perturbation with a variety of step length–step time combinations, ranging from a slowly executed large step, to a quick small step. Previous studies (Hsiao and Robinovitch, 1999) suggest that one should be able to recover from a specific perturbation with a variety of step length–step time combinations, ranging from a more slowly executed large step, to a quicker small step. These same considerations suggest that recovery ability should associate independently with both step length and step time.

In the current study, we sought improved understanding of these relations, by testing the following hypotheses: (1) recovery ability depends on step length; (2) recovery ability associates with step time, when step length is held constant; and (3) recovery ability is different in young and elderly women, when step length is held constant. To test these hypotheses, we conducted experiments with young and elderly women, to measure the maximum release angle where they could recover balance by taking a single forward step, onto floor-mounted targets located at specific forward distances from the toes.

We found that increases in step length caused a large improvement for both young and elderly women in the maximum release angle where balance recovery was possible. However, at a given step length, the maximum release angle was larger for young women than elderly women. This appeared to be due to their faster step lengths, and ability to generate larger torques in the lower extremity joints leg during step contact.

2. Methods

2.1. Participants

We conducted balance recovery experiments with ten healthy, community-dwelling elderly women (mean age = 75 (SD 3) yrs, age range: 71–83 yrs; mean body mass = 65 (SD 14) kg; mean height = 1.57 (SD 0.09) m) and ten young women (mean age = 28 (SD 4) yrs, age range: 18–32 yrs; mean body mass = 62 (SD 10) kg; mean height = 1.63 (SD 0.07) m). Potential participants were excluded if they had diagnosed neurological disease (Parkinson's Disease, peripheral neuropathy, or stroke), gait impairments, severe arthritis, required a cane or walker for ambulation, had major uncorrected visual deficits, or cognitive impairment as defined by a score of less than 27 on the mini-mental state exam (Folstein et al., 1985). We also excluded individuals who used medications that have a documented effect on balance or risk for falls (sedatives, tranquilizers, antiarrhythmics, antihypertensives, or antidepressants). We obtained informed written consent from each participant, and the experiment was approved by the Institute Review Board at Simon Fraser University and the University of California, San Francisco.

2.2. Experimental protocol

We conducted four series of balance recovery trials with each participant. In each series, we measured the maximum initial lean angle where the participant could be released, and recover balance by taking a single forward step. To conduct a trial, we inclined the participant into a stationary

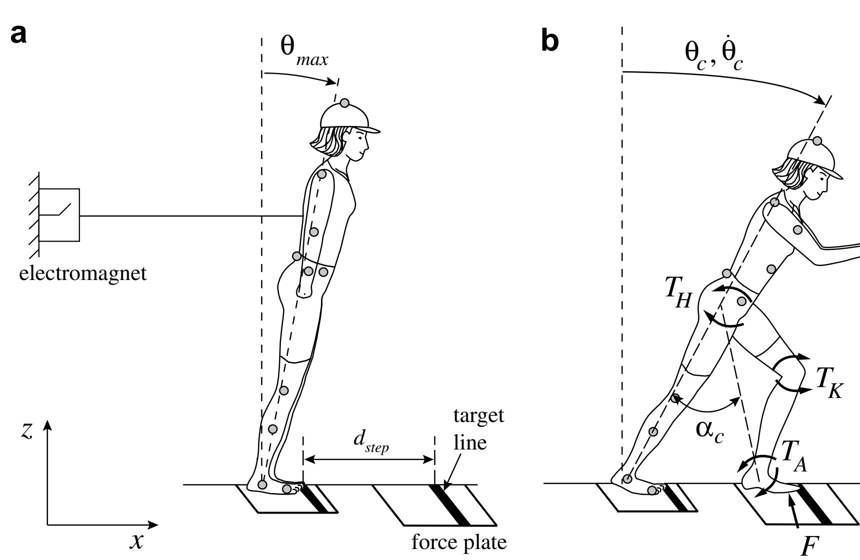

Fig. 1. Balance recovery experiment. (a) We used a horizontal tether and electromagnet to determine the maximum release angle (θ_{max}) where the participant could be released and recover balance by taking a single forward step. In three of the series, the participant stepped to a target line, located a distance (d_{step}) from the toes of 15%, 25%, or 35% body height. In *Max Step Length* trials, the target line was removed and the participant recovered with her "largest and fastest step." (b) We used a force plate to measure contact forces (*F*), and calculated sagittal-plane torques at the ankle, knee, and hip (T_A, T_K, T_H) from inverse dynamics. We also measured the stepping angle (α_c) and body lean angle and velocity (θ_c, $\dot{\theta}_c$) at the instant of step contact. Joint torque was defined positive if plantarflexor at the ankle, and extensor at the knee and hip.

forward leaning position via a horizontal tether that attached at one end to an electromagnet and at the other end to a chest harness worn by the participant (Fig. 1). We then instructed the participant that their goal was to recover balance with a single forward step once the tether was suddenly released (by discharging the electromagnet; brake release time = 15 ms). The participant was also instructed that, after stepping, she should remain stationary for approximately three seconds and then resume an upright stance. We considered the recovery attempt to be successful if the participant was able to complete this instruction without taking a second step.

In three of the four series, we controlled the length of the step by instructing the participant to step to a floor-mounted target (a 2 cm wide strip of tape) located a horizontal distance of 15%, 25%, or 35% of body height from the anterior edge of the toes. We selected these step lengths because (a) they are within the range measured previously in balance recovery studies (Luchies et al., 1994; Thelen et al., 1997; Wojcik et al., 1999; Hsiao and Robinovitch, 1999; Medell and Alexander, 2000; Mcilroy and Maki, 1996), (b) we wished to explore how recovery ability varies over the a large range in step lengths, and (c) they were found in our preliminary trials to be feasible for our older female participants. In the final series, we instructed the participant to recover balance by taking "as large and as fast a step as possible." We refer to the four series as 15% *BH*, 25% *BH*, 35% *BH*, and *Max Step Length*, respectively, where *BH* refers to "body height."

During each trial, we recorded ground reaction forces under the stepping foot with a force plate (model 6090H, Bertec Corp., Worthington, OH, USA) mounted flush with the walkway and concealed under a thin layer of linoleum. We also used a six camera motion measurement system (MacReflex, Qualisys Inc., Glastonbury, CT, USA) to record the positions of 25 mm diameter reflective spherical markers secured (with double-sided tape) at the ankles (lateral malleoli), toes (fifth metatarsals), mid-shanks, knees (lateral femoral epicondyles), mid-thighs, anterior superior iliac spines, sacrum (L5/S1 junction), wrists, elbows (lateral humeral epicondyles), shoulders (acromion processes), and crown of the head. During the trials, the participant was barefoot and wore tight-fitting Spandex short pants and sports bra to minimize marker occlusion. We acquired force data at 960 Hz and motion data at 60 Hz. The measurement accuracy of the camera system was approximately 1.0 mm. We filtered marker position and force plate data with a recursive, fourth-order lowpass Butterworth filter having cut-off frequencies of 6 Hz and 96 Hz, respectively. We acquired data for 4 s, beginning approximately 1 s before tether release. For safety, we secured the participant to a fall restraint harness, which was slack during stepping and independent from the horizontal tether.

For each participant, the first trials involved lean angles of approximately two degrees, where they could recover balance easily. We then iteratively adjusted the length of the tether until we determined the maximum initial lean angle (with a resolution of 5 mm in tether length, and approximately 0.2° in lean angle) beyond which the participant was unable to recover balance in two consecutive trials. Once the maximum release angle was identified, we conducted two additional trials at that lean angle. During the trials, we inserted a random time delay of 1–10 s between the "ready" signal and tether release. Therefore, while the participant knew the direction and magnitude of the perturbation, they were unable to predict the instant of release. We repeated trails if the step length obviously exceeded the target line. We classified trials as failed attempts if the participant required support from the fall restraint harness.

Before the instant of tether release, the participant stood with her feet shoulder-width apart, knees and hips extended, and arms at her sides. No restrictions were placed on arm usage after tether release. Rest breaks of approximately 2 min duration were provided between successive trials. For all the participants, the series were presented in order of decreasing step length (*Max Step Length*, 35% *BH*, 25% *BH*, and 15% *BH*). We found in preliminary experiments that this was preferable to a randomized presentation, since it (a) allowed the participant to select a maximum step length without the confounding influence of previous trials, and (b) reduced the length of the testing session and thereby helped to prevent participant fatigue. Despite the demanding nature of the experiment, all participants were able to complete the experiment, with no resulting injuries.

2.3. Data analysis

We determined several kinematic parameters from each trial. We calculated the body lean angle $\theta(t)$ as the angle from the vertical to a line connecting the midpoint of the two lateral malleolus markers to the midpoint of the two acromium markers (Fig. 1). For each of the four trials acquired at the participant's maximum (recoverable) initial lean angle, we calculated θ_{max} as the average value of $\theta(t)$ over the 500 ms interval preceding tether release. We defined step contact time t_{step} as the time interval between tether release and the instant the force plate (which the participant stepped upon) recorded a force greater than 10 N.

We also determined several kinetic parameters related to the contact phase of the step. We used a bottom-up inverse dynamics routine (Cappozzo et al., 1975) to estimate the rates of development and peak magnitudes of lower extremity joint torques during step contact. We focused on torques in the sagittal plane (ankle plantarflexion/dosiflexion, and knee and hip flexion/extension), which would appear to have a much more dominant role than internal/external or abduction/adduction in halting the body's forward motion. In this analysis, we estimated the ankle and knee joint centers from markers on the lateral malleolus and lateral femoral condyle, respectively, and the hip joint center from the anterior superior iliac spine and sacral markers using a technique proposed first by Vaughan et al. (1992). We then calculated joint angular displacements, velocities, and accelerations based on marker and joint centre coordinates in the global x–z plane, which was aligned closely with the participant's sagittal plane.

2.4. Statistics

We used a two factor ANOVA to determine whether the maximum release angle (θ_{max}) associated with step length (a repeated-measures factor with four levels) and age (a grouping factor with two levels). Where applicable, we conducted post-hoc comparisons with t-tests. Finally, we determined whether correlations exist between θ_{max} and t_{step}, F_{max}, and peak magnitudes and rates of lower extremity torque generation during stepping. In all tests, we regarded $P < 0.05$ to indicate statistical significance. All analyses were performed with the SPSS statistical software package (SPSS Inc., Chicago, IL, USA), and were based on average parameter values (from the four repeated trials) for each participant and experimental series.

3. Results

We found that θ_{max} associated with both step length ($P < 0.001$) and age ($P < 0.001$; Table 1). Furthermore, there was a significant interaction between step length and age ($P < 0.001$). Post-hoc analysis indicated that θ_{max} was significantly different at each of the four step lengths. When the step length increased from 15% *BH* to 25% *BH*, the maximum release angle increased on average by 36% or 4.1° on average (from 11.4° (SD 2.6) to 15.5° (SD 3.6); SE of the difference = 0.38°; 95% CI for the difference = 2.9–5.2°; $P < 0.001$; Table 1). The mean increase was 31% in elderly and 36% in young participants. When the step length increased from 25% *BH* to 35% *BH*, θ_{max} increased by 15% or 2.4° on average (from 15.5° (SD 3.6) to 17.9° (SD 5.0); SE = 0.57°; 95% CI = 0.8–4.1°; $P = 0.003$). The mean increase was 6% in elderly and 23% in young. Finally, when the step length increased from 35% *BH* to *Max Step Length*, θ_{max} increased by 23% or 4.2° on average (from 17.9° (SD 5.0) to 22.1° (SD 6.0); SE = 0.58°; 95% CI = 2.5–5.9°; $P < 0.001$). The mean increase was 17% in elderly and 27% in young.

For all but the 15% *BH* condition, there was significant correlation between θ_{max} and variables representative of step kinematics and step kinetics (Table 2). For example, in the 35% *BH* condition, θ_{max} decreased with increasing t_{step} ($r = -0.74$; $P < 0.001$), and increased with increasing F_{max} ($r = 0.84$; $P < 0.001$), increasing peak ankle plantarflexor torque ($r = 0.78$; $P < 0.001$) and increasing peak hip extensor torque ($r = 0.78$; $P < 0.001$). There was no correlation between θ_{max} and peak knee extensor torque, and generally little correlation between θ_{max} and rates of torque development.

At a given step length, young participants recovered from larger magnitudes of θ_{max} than elderly, and as step length increased, so did the difference between θ_{max} in elderly and young (Table 1). At 15% *BH*, the difference between mean values of θ_{max} in elderly and young was 21% or 2.2° (10.3° (SD 1.8) versus 12.5° (SD 2.9); SE of the difference = 1.1°; 95% CI for the difference = −0.5° to 4.5°; $P = 0.055$). At 25% *BH*, the difference was 30% or 3.9° (13.5° (SD 1.8) versus 17.5° (SD 3.8); SE = 1.3°; 95% CI = 1.1–6.7°; $P = 0.009$). At 35% *BH*, the difference was 51% or 7.3° (14.3 (SD 2.7) versus 21.6° (SD 4.0); SE = 1.5°; 95% CI = 4.0–10.5°; $P < 0.001$). Finally, in the *Max Step Length* condition, the difference was 63% or 10.7° (16.8° (SD 2.5) versus 27.5° (SD 2.5); SE = 1.1°; 95% CI = 8.3–13.0°; $P < 0.001$).

There were significant differences between young and elderly in both step kinematics and kinetics (Table 1 and Fig. 2). Young participants used smaller values of t_{step},

Table 1
Mean recovery angles and step characteristics (with one standard deviation shown in parentheses)

Parameter	Step length							
	15% *BH*		25% *BH*		35% *BH*		*Max Step Length*	
	Elderly	Young	Elderly	Young	Elderly	Young	Elderly	Young
θ_{max} (deg)	10.3 (1.8)	12.5 (2.9)	13.5 (1.8)	17.5 (3.8)	14.3 (2.7)	21.6 (4.0)	16.8 (2.5)	27.5 (2.5)
t_{step} (ms)	390 (60)	320 (30)	410 (40)	350 (30)	480 (60)	380 (20)	490 (30)	440 (40)
d_{step} (% *BH*)	17 (2)	18 (1)	26 (2)	28 (1)	36 (3)	38 (2)	44 (5)	63 (6)
α_c/θ_c	1.17 (0.15)	1.15 (0.17)	1.37 (0.15)	1.27 (0.13)	1.68 (0.25)	1.38 (0.14)	1.73 (0.11)	1.52 (0.18)
F_{max} [a]	11.19 (1.80)	11.87 (1.79)	11.85 (2.22)	13.75 (2.37)	10.67 (1.66)	14.07 (2.16)	11.40 (2.65)	11.77 (2.41)
Ankle:								
Peak plantarflexor torque [b]	0.80 (0.52)	0.91 (0.40)	0.66 (0.17)	0.95 (0.25)	0.53 (0.17)	0.89 (0.24)	0.54 (0.14)	0.80 (0.30)
Dev. rate [c]	6.6 (4.9)	7.5 (2.6)	3.4 (2.0)	8.0 (2.9)	3.7 (1.9)	7.4 (2.4)	3.4 (2.0)	5.3 (1.7)
Knee:								
Peak extensor torque [b]	0.13 (0.21)	0.04 (0.27)	0.22 (0.24)	0.15 (0.25)	0.37 (0.24)	0.20 (0.26)	0.48 (0.22)	0.55 (0.33)
Dev. rate [c]	2.4 (1.9)	2.2 (2.7)	5.9 (3.2)	2.7 (1.7)	5.0 (2.2)	3.6 (2.7)	5.9 (3.2)	8.1 (5.4)
Hip:								
Peak extensor torque [b]	0.75 (0.48)	0.97 (0.34)	0.88 (0.45)	1.29 (0.34)	0.69 (0.21)	1.48 (0.38)	0.86 (0.31)	1.65 (0.36)
Dev. rate [c]	6.6 (3.4)	10.8 (4.1)	8.8 (6.2)	15.1 (5.5)	6.3 (3.3)	19.7 (5.8)	8.8 (6.2)	22.6 (8.8)

[a] Peak forces are normalized by body mass (and thus, have units of N/kg).
[b] Peak torques are normalized by the product of body mass* body height (and thus, have units of N m/(kg m)).
[c] Torque development rates are normalized by the product of body mass* body height (and thus, have units of N m/(kg m s)).

Table 2
Pearson correlation coefficients and significance levels indicating how the maximum release angle (θ_{max}) correlated with step kinematics and kinetics

Parameter	Step length			
	15% BH	*25% BH*	*35% BH*	*Max Step Length*
t_{step} (ms)	$r = -0.337$	$r = -0.567^{B}$	$r = -0.738^{B}$	$r = -0.617^{B}$
	$P = 0.146$	$P = 0.009$	$P < 0.0005$	$P = 0.004$
F_{max}[a]	$r = 0.325$	$r = 0.643^{B}$	$r = 0.842^{B}$	$r = 0.209$
	$P = 0.175$	$P = 0.002$	$P < 0.0005$	$P = 0.376$
Ankle:				
Peak plantarflexor torque[b]	$r = 0.262$	$r = 0.744^{B}$	$r = 0.782^{B}$	$r = 0.550^{A}$
	$P = 0.278$	$P < 0.0005$	$P < 0.0005$	$P = 0.012$
Development rate[c]	$r = -0.390$	$r = -0.479^{A}$	$r = -0.418$	$r = -0.504^{A}$
	$P = 0.098$	$P = 0.032$	$P = 0.067$	$P = 0.023$
Knee:				
Peak extensor torque[b]	$r = -0.087$	$r = -0.256$	$r = -0.082$	$r = 0.241$
	$P = 0.724$	$P = 0.276$	$P = 0.731$	$P = 0.306$
Development rate[c]	$r = 0.275$	$r = -0.010$	$r = 0.358$	$r = -0.385$
	$P = 0.254$	$P = 0.967$	$P = 0.122$	$P = 0.094$
Hip:				
Peak extensor torque[b]	$r = 0.320$	$r = 0.651^{B}$	$r = 0.779^{B}$	$r = 0.658^{B}$
	$P = 0.182$	$P = 0.002$	$P < 0.0005$	$P = 0.002$
Development rate[c]	$r = 0.128$	$r = -0.069$	$r = -0.240$	$r = -0.496^{A}$
	$P = 0.601$	$P = 0.773$	$P = 0.307$	$P = 0.026$

[A] Correlation is significant at the 0.05 level.
[B] Correlation is significant at the 0.01 level.
[a] Peak forces are normalized by body mass (and thus, have units of N/kg).
[b] Peak torques are normalized by the product of body mass* body height (and thus, have units of N m/(kg m)).
[c] Torque development rates are normalized by the product of body mass* body height (and thus, have units of N m/(kg m s)).

larger values of F_{max}, and larger magnitudes and rates of development in ankle plantarflexor and hip extensor torque during step contact.

4. Discussion

We examined how step length affected young and elderly participants' ability to recover a stable upright stance with a single step. We found that, by increasing the length of their step, participants were able to substantially increase the maximum release angle where balance recovery was possible. For example, when step length increased from 15% to 25% of body height, average values of θ_{max} increased by 36% in young and 31% in elderly participants. These differences are functionally important and well above the measurement accuracy of our experiment.

We observed that, for a given step length, the maximum release angle tended to correlate with step contact time, peak contact force, peak ankle plantarflexor torque, and peak hip extensor torque. An exception was in the 15% *BH* condition, where these trends failed to reach statistical significance, perhaps due to our small sample size. Peak knee extensor torque did not correlate with the maximum release angle, and tended to be smaller in young than elderly women. This supports the notion that the leg is stabilized during step contact primarily through eccentric torques at the ankle and hip, and that relatively low stiffness (and greater flexion) at the knee may be advantageous.

We also found that, for a given step length, young participants were able to recover from larger release angles than elderly participants, and that the difference in recovery ability increased with increases in step size (from 21% for a step size of 15% *BH*, to 51% for a step size of 35% *BH*). This seemed attributable to quicker step contact times (and therefore shorter descent times) and greater peak contact forces and magnitudes of ankle plantarflexor and hip extensor torque during step contact. In the 25% *BH* and 35% *BH* conditions, young women tended to overshoot the target line slightly more than elderly (by about 2% of *BH* on average) and this may have also contributed to their improved recovery ability.

Our results complement previous studies that examined age differences in ability to recover balance with a single forward step when there is no restriction on step length. For example, Wojcik and co-workers (1999) found that average values of the maximum recovery angle were 89% greater in young than in elderly women (30.7° ± 2.8° versus 16.2° ± 4.5°). This seemed to associate with elderly participants taking smaller and slower steps than young: average step lengths were 67% *BH* in young and 55% *BH* in elderly, and step contact times averaged 462 ms in young and 600 ms in elderly. When compared to our results, young women in Wojcik et al.'s study had step lengths and step contact times that were similar to ours for the *Max Step Length* condition, but their elderly women took slower, larger steps than ours. Thelen and co-workers (1997) found that average maximum recovery angles were 36% greater

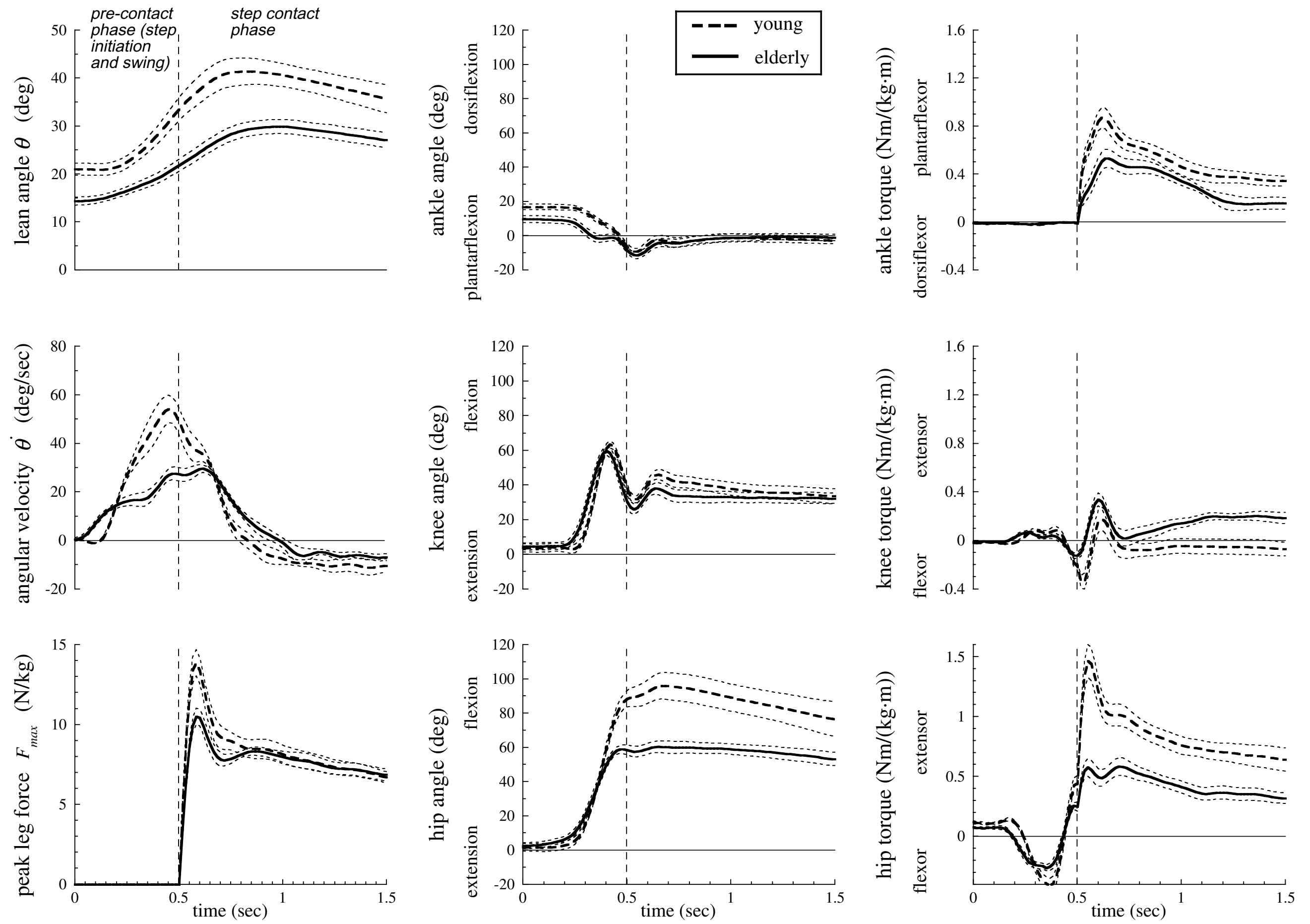

Fig. 2. Ensemble-averaged stepping behaviour for elderly and young participants at 35% *BH* step length. Data are synchronized to the instant of step contact, at $t = 0.5$ sec, and shown by the vertical dashed line (therefore, $t = 0$ is not the instant of tether release, but instead 0.5 s before step contact). Heavy solid lines show mean elderly behaviour, and heavy dashed lines show mean young behaviour. Surrounding dotted lines show one standard error.

in young than in elderly men (32.5° ± 4° versus 23.9° ± 4°). They observed average step lengths of 69% *BH* in their young men and 51% *BH* in their elderly men, and average step contact times of 462 ms in young and 472 ms in elderly men. Our findings support their interpretation that between-group differences were due largely to differences in step length. Grabiner et al. (2005) used a tether-release protocol to examine the ability of elderly participants (of mean age 72 (SD 5) yrs) to recover balance with a single step, when no restrictions were placed on step length. They observed a maximum recovery angle for women (n=36) averaging 14.9° (SD = 3.0), which is similar to our average value at the 35% *BH* condition, but less than our average in the *Max Step Length* condition. They found that about 30% of the variance in recovery ability was accounted for by lower extremity strength, most noticeably isokinetic ankle dorsiflexor strength, which contributes to adequate ground clearance during the swing phase of stepping. Step lengths or step contact times were not reported in their analysis. King et al. (2005) examined the effect of step length on the mechanics of balance recovery in healthy young men. They found that, while step length had no effect on the time required for step initiation (as observed by Do et al. (1982)), larger steps were associated with longer swing phase durations and larger impulses between the foot and the ground at landing. We observed similar trends, and extend King's findings to show that the net effect of increasing step length is to increase the ability of both young and elderly women to recover balance.

There are limitations and strengths to our approach of instructing participants to "recover balance with a single step." On the one hand, previous studies have shown that, for the majority of participants, the natural response to regain balance following an unexpected perturbation (elicited by more general instructions such as "recover balance" or "prevent a fall") is to take two or more steps (Hsiao and Robinovitch, 1998, 2001; Mcilroy and Maki, 1996). For such individuals, a more effective approach to balance training may be to enhance the multiple-step response, instead of replacing it with the single-step response we studied. On the other hand, the presence of obstacles – such as walls, furniture, or curbs – often renders a multiple-step response infeasible, and necessitates that a single step be used to recover balance. Furthermore, each step of a multiple-step response can be regarded as an attempt to recover balance, albeit under changing initial conditions. Accordingly, we expect that similar biomechanical variables – related to step length, step contact time, and peak attainable hip extensor and ankle plantarflexor torque – will govern the effectiveness of single step and multiple-step recovery attempts. Finally, our approach allowed us to determine the biomechanical factors that govern performance on a specific measure of postural stability – ability to recover balance with a single forward step. This would be impossible to achieve with a protocol that placed no restrictions on the number or characteristics of steps – especially since we have little understanding of whether the number of steps used to recover balance reflects how "good" one's postural stability is, or whether additional steps are used more often out of preference or necessity.

There are three additional limitation of our study worth mentioning. First, some participants' performance may have been affected by fatigue, motivation, or fear. However, we attempted to minimize these potential confounders by providing frequent rest periods, strong verbal encouragement, and by working to ensure that participants were comfortable with the safety precautions in our experiment. Second, we conducted the experimental series in order of decreasing step length. This may have introduced a slight learning effect favoring performance at small step lengths, or a slight fatigue effect acting to produce an opposite trend. However, given the constant rest breaks provided to participants and the nature of our performance task, we regard each of these effects as minimal. Third, we did not measure or include step width in our analysis. However, ability to recover balance from the forward fall induced by tether release requires stabilization of the whole-body centre-of-gravity in the medial–lateral as well as the anterior–posterior direction, and will therefore depend on step width as well as step length. Furthermore, when compared to young, elderly participants exhibit wider and more variable step widths during walking (Owings and Grabiner, 2004), and similar trends may contribute to age-related differences in recovery ability in the tether-release protocol.

In conclusion, we found that, for both young and elderly women, there is a dramatic increase in ability to recover balance with a single step when the length of the step is increased. We also found that at a given step length, young participants were able to recover balance from larger recovery angles than the elderly. This appeared to be attributable to shorter step contact times and larger ankle plantarflexor and hip extensor torques in the stepping leg during step contact. These results suggest that exercise-based programs to enhance postural stability in the elderly should focus on training elderly participants to take large steps, and enhancing those components of strength, reaction time, and flexibility that are essential to this capacity.

Acknowledgement

This study was partially funded by an operating grant from the Canadian Institutes for Health Research (MOP 64414).

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