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Are age-related impairments in change-in-support balance reactions dependent on the method of balance perturbation?

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ABSTRACT

Rapid "change-in-support" (stepping or grasping) balance-recovery reactions play a critical role in preventing falls. Studies investigating age-related impairments in these reactions using differing perturbation methods have shown contradictory results. The discrepancies could be due to the different mechanical and sensory stimuli provided by the different perturbation methods, but could also be due to other confounding factors (e.g. differences in perturbation predictability). This study compared two commonly used perturbation methods: weight-drop cable-pulls (CPs) and motor-driven surface-translations (STs). For each perturbation method, effects of aging on the change-in-support reactions were established by comparing 10 young (22–28 years) and 30 older (64–79 years) adults, using large unpredictable multi-directional perturbations similar to those used in previous studies showing age-related differences. Age-related differences in the pattern and spatio-temporal features of the limb movements were examined for stepping and grasping reactions evoked by antero-posterior perturbation of stance, as well as stepping reactions evoked by lateral perturbations delivered while subjects walked "in-place". Although age-group effects were almost always more pronounced for ST perturbations, the direction of the effect was always the same for both perturbation methods; hence, the perturbation-dependent differences in mechanical and sensory stimuli did not seem to be a critical factor. Perturbation waveform appeared to be a more important factor. For the perturbation methods used here, the ST perturbations were more destabilising than the CP perturbations (leading to a more rapid rise in perturbatory ankle-torque and greater centre-of-mass motion prior to the onset of the postural reaction), and were consequently more effective in revealing age-related deficiencies.

1. Introduction

The ability to react to sudden perturbations is critical to balance control. Of particular importance in preventing falls are change-in-support reactions, involving rapid stepping and grasping movements (Maki and McIlroy, 2006). These reactions are the only defence against large postural perturbations (Shumway-Cook and Wollacott, 1995), but are frequently recruited following smaller perturbations when subjects are allowed to react naturally (McIlroy and Maki, 1993b; Jensen et al., 2001).

Age-related differences in change-in-support reactions have been studied using cable-pull (CP, e.g. Luchies et al., 1994; Rogers et al., 2001), surface-translation (ST, e.g. McIlroy and Maki, 1996; Brauer et al., 2002), and release-from-lean (RFL) perturbations (e.g. Thelen et al., 1997; Hsiao-Wecksler and Robinovitch, 2007). Contradictory age-related effects have emerged from these studies. For example, some studies showed that foot-off times were slower in older adults (OA), some showed that young adults (YA) were slower, and some showed no age-related difference (Table 1).

The cause of the contradictory findings is unclear. One possibility pertains to differences in perturbation method, which result in differing mechanical and sensory stimuli. For example, CPs apply pressure at the pelvis whereas STs induce shear forces at the foot–sole; therefore, there are differences in cutaneous stimuli. Furthermore, differences in the point-of-application of perturbatory force could affect induced patterns of motion as well as associated proprioceptive, visual and vestibular stimuli (Liu et al., 2003). Differing mechanical and sensory stimuli, and age-related differences in the ability to respond to specific types

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Table 1
Examples of previous perturbation studies showing contradictory age-related differences in characteristics of stepping reactions.

Study	Unpredictability			Instruction	AP-step measures			ML-step measures	
	Onset timing	Magnitude	Direction[a]		Foot-off time	Swing duration	Step length	Cross-over steps	Foot collisions
Cable-pull (CP) perturbations:									
Luchies et al. (1994)	Yes	Yes (10 different pull distances)	No (B only)	Not specified	OA<YA[b]	OA<YA	OA<YA	–	–
Rogers et al. (2001)[c]	Not specified	Yes (five different magnitudes)	No (F only)	React naturally	OA<YA	OA<YA	–	–	–
Rogers et al. (2003)[c]	Not specified	No	No (F only)	React naturally	OA = YA	–	–	–	–
Schulz et al. (2005)	Not specified	Yes (five different magnitudes)	Yes (F, B)	React naturally	OA = YA	OA = YA	OA<YA (B only)	–	–
Mille et al. (2005)[c]	Not specified	No	Yes (L, R)	React naturally	–	–	–	OA>YA	OA>YA
Present findings	**Yes**	**Yes (STs included)[d]**	**Yes (L, R, F, B)**	**React naturally but minimise number of steps**	OA<**YA**	**OA = YA**	**OA<YA**	**OA = YA[e]**	**OA = YA[e]**
Surface-translation (ST) perturbations:									
McIlroy and Maki (1996)	Yes	No	Yes (F, B)	Try not to fall	OA = YA	OA = YA	OA = YA	–	–
Maki et al. (2000)	Yes	Yes ('low' and 'high' magnitude)[d]	Yes (L, R, F, B)	React naturally	–	–	–	OA = YA[e]	OA>YA[e]
Present findings	**Yes**	**Yes (CPs included)[d]**	**Yes (L, R, F, B)**	**React naturally but minimise number of steps**	OA<**YA**	**OA = YA**	**OA = YA**	**OA = YA[e]**	**OA>YA[e]**
Release-from-lean (RFL) perturbations:									
Thelen et al. (1997)	Yes	No	No (F only)	Take a single step forward with right foot	OA>YA	–	–	–	–
Wojcik et al. (1999)	Yes	No	No (F only)	Take a single step forward with right foot	OA>YA	–	OA>YA	–	–

The findings of the present study are also included (bold characters).
[a] Direction of falling motion induced by perturbation: L = leftward, R = rightward, F = forward, and B = backward.
[b] OA = older adults ($\geqslant$65 years), YA = young adults ($\leqslant$40 years). "OA<YA" indicates that the OA mean was significantly smaller than the YA mean, etc.
[c] These cable-pull studies used an electromechanical actuator, rather than a weight-drop mechanism.
[d] These studies included ST perturbations that were also unpredictable in terms of their waveform, i.e. the timing of the acceleration and deceleration of the moving surface was not the same for all trials.
[e] These findings pertain to trials in which the perturbation was delivered while subjects walked "in-place" (in all other cases the perturbations were applied during bipedal stance).

of stimuli, may influence characteristics of balance-recovery reactions and the degree to which age-related differences are observed. It is also possible that differences in the time-history and amplitude of the perturbatory force affect the degree to which age-related differences emerge; however, few studies have provided details regarding the perturbation waveform and reported amplitude variables can be difficult to compare (e.g. cable-pull force versus support-surface acceleration).

The contradictory findings could also be due to differences in the predictability of perturbation characteristics, which could affect the ability to adopt predictive control strategies (Horak et al., 1989; Maki and Whitelaw, 1993). During release-from-lean perturbations, perturbation direction and magnitude are entirely predictable. Unpredictable multi-directional CP and ST perturbations are possible (Henry et al., 1998; Luchies et al., 1999; Maki et al., 2000; Mille et al., 2005; Schulz et al., 2005); however, CP and ST studies have varied in the degree of unpredictability used. Additionally, there are often differing instructions given to subjects, which can have a strong influence on certain features of postural reactions (Maki and McIlroy, 1997).

This study aimed to determine if previously reported age-related differences in change-in-support reactions are dependent on perturbation method, under conditions where other confounding factors are controlled. We compared CPs delivered by a weight-drop apparatus and STs delivered by a motor-driven motion-platform, using perturbation parameters (weight-drop magnitude/distance, platform-acceleration profile) similar to previous studies (e.g. Luchies et al., 1994; McIlroy and Maki, 1996). In each case, we gave the same instructions to subjects and varied the perturbation features in an unpredictable manner. We hypothesised that both perturbation methods would reveal the same fundamental age-related deficiencies. However, we also suspected that the two methods would exhibit differences in perturbation waveform that could influence the degree to which these deficiencies are revealed. To explore this possibility, we developed a simple model to compare the time-history of the perturbatory torque and also analysed differences in evoked center-of-mass (COM) motion.

2. Methods

We recruited 10 YA (22–28 years; five men; height 1.63–1.83 m; weight 57–104 kg) and 30 community-dwelling OA with a history of falls or instability (64–79 years; 15 men; height 1.51–1.82 m; weight 52–118 kg). Subjects were right handed with no neuromusculoskeletal conditions adversely affecting daily activities. Ethical approval was obtained from the institutional review board and subjects provided written informed consent. The OA were participants in a balance-training study; the pre-training data presented here are also reported as part of that study (Mansfield et al., 2008).

As detailed previously (Mansfield et al., 2007), subjects either stood or walked in-place on a large (2×2 m) multi-axis motion platform that delivered ST perturbations (Fig. 1). Multi-axis CP perturbations were delivered using a weight-drop system connected to a belt (worn at the height of the anterior–superior iliac spines) via

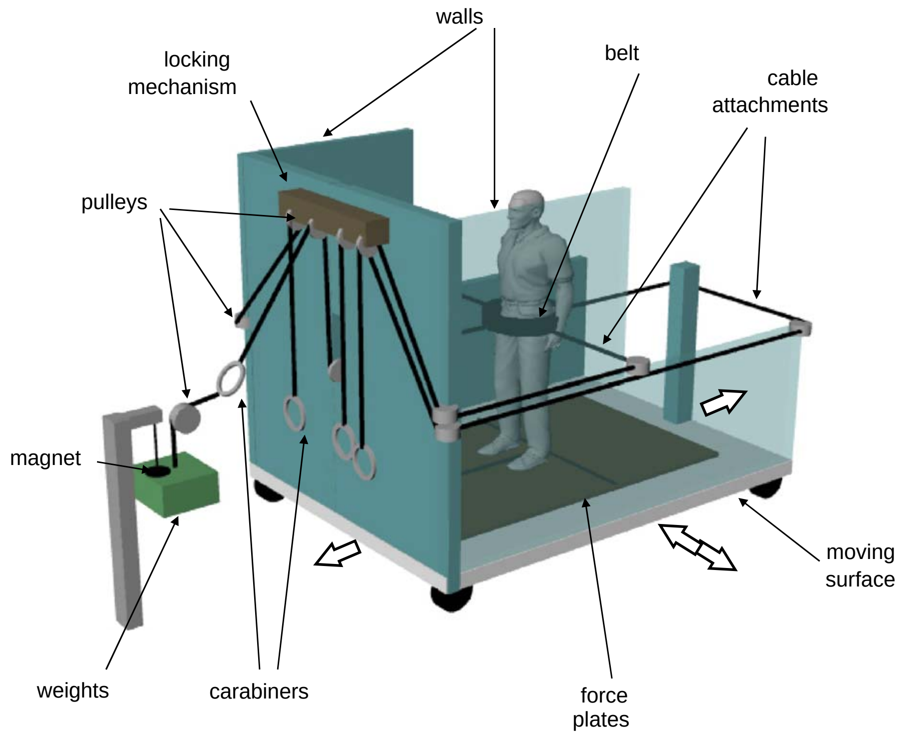

Fig. 1. Experimental set-up for delivering the surface-translation (ST) and cable-pull (CP) perturbations. Unpredictable multi-directional (forward, backward, left or right) STs were delivered via rapid horizontal translation of the computer-controlled motor-driven motion platform. Unpredictable multi-directional CPs were applied by dropping a weight attached to a pelvic belt (worn at the level of the anterior–superior iliac spines) via a cable and pulley system, using a computer and electromagnet to control the timing of the weight drop. Four cables were attached to the belt and the experimenter connected the carabiner for one of these cables to the weight-drop mechanism, prior to each CP-trial, in order to vary the CP direction unpredictably (in ST-trials, none of the cables were attached to the weight). A locking mechanism provided an equal amount of slack (~2–4 cm) in each of the four cables and thereby ensured that subjects could not detect which (if any) of the cables were attached to the weight, prior to the release of the locking mechanism, which occurred immediately prior to the onset of the CP or ST perturbation. In grasping trials, a cylindrical handrail (height: 55% subject height, diameter: 38 mm, length: 1.63 m) was mounted on the platform, to the right of the subject (25% subject height from the midline of the subject's body), and foam blocks (40 cm high) were placed around the feet in order to deter stepping. To deter arm reactions in stepping trials, subjects held a lightweight rod behind their back (Maki et al., 2000), and were instructed not to move their arms or release the rod. At the start of each trial, subjects either stood or walked "in-place" at the centre of the platform, with the feet in a comfortable standardized position (McIlroy and Maki, 1997).

cables and pulleys. The CP system was mounted on the motion platform, allowing perturbation method, direction and timing to be varied unpredictably from trial-to-trial. After 12 initial familiarisation trials, subjects completed three trial blocks focussing on: (1) stepping evoked by antero-posterior (AP) perturbation of stance; (2) stepping evoked by medio-lateral (ML) perturbation while walking in-place; (3) grasping reactions evoked by backward perturbation of stance (Table 2). The walk-in-place task was used in ML-step trials because previous work has shown that this task exacerbates age-related problems in avoiding collision between the swing and stance limbs (Maki et al., 2000).

Three force-plates recorded ground-reaction forces (Fig. 1), a three-dimensional motion-capture system recorded coordinates of markers on the feet and arms, and four video cameras recorded gross motor behaviours. Force-sensing resisters recorded handrail-contact and load cells recorded the CP force and safety-harness loading. Accelerometers and a linear potentiometer measured motion-platform acceleration and displacement. The accelerometer signals were used to correct force-plate measures for inertial artifacts arising from platform motion (Maki, 1987). Surface electromyographic (EMG) activity was recorded bilaterally from tibialis anterior, medial gastrocnemius, medial deltoid, and biceps brachii. All signals were low-pass filtered (10 Hz) and sampled at 200 Hz, except the video (sampled at 60 Hz) and EMG (band-pass filtered 10–500 Hz, sampled at 1000 Hz).

To compare the destabilising effect of CP and ST perturbations, COM displacement and velocity were calculated for AP-perturbation trials (trial-block #1), by integrating the net AP-force acting on the body (Fig. 2). An inverted-pendulum model estimated the perturbatory ankle-torque resulting from each perturbation method (Fig. 3).

Outcome measures were those used in previous studies examining age-related differences in change-in-support reactions. AP- and ML-step reactions (trial-blocks #1 and #2, respectively) were characterised by: frequency of multi-step reactions, stepping pattern, and frequency of arm reactions (despite instructions not to move arms). We also determined the frequency of "extra" lateral steps (AP-perturbations, trial-block #1) and frequency of collisions between the swing and stance limbs (ML perturbations, trial-block #2). All of the above were detected from the kinematic data and confirmed by inspection of the videos. For the initial step in each trial (AP-perturbations, trial-block #1), we also analysed: (1) onset of preceding ankle-muscle activation (the "automatic postural response"; (Nashner and Cordo, 1981)); (2) foot-off and foot-contact times; (3) occurrence of "anticipatory postural adjustments" (APAs, defined by an initial lateral centre-of-pressure excursion >4 mm toward the swing limb prior to unloading; (McIlroy and Maki, 1999)); (4) step length and width. For grasping trials (trial-block #3), we analysed biceps and medial-deltoid latencies, and handrail-contact time. Harness-assisted recoveries (harness loading >20% body weight) were analysed for all trials. Timing measures were determined relative to perturbation onset (surface acceleration $>0.1\,m/s^2$; cable force >5 N). EMG onset latencies were determined by a computer algorithm (McIlroy and Maki, 1993a) and confirmed by visual inspection.

Data from each perturbation method were first examined separately, using a one-way repeated measures ANOVA to determine if there were significant age-group effects. The strength of the age-effect was quantified by calculating the "effect size" (i.e. difference in OA and YA means divided by pooled standard deviation). Two-way repeated measures ANOVA, with age-group and perturbation method as factors, were then used to examine the age-by-method interaction to determine if the age-effect was dependent on perturbation method. Data were rank-transformed prior to analysis, to avoid errors arising from violation of assumptions underlying the ANOVA (Conover and Iman, 1981).

3. Results

The model revealed a more rapid rise in the initial perturbatory ankle-torque generated by STs, compared to CPs, although the

Table 2
Details of the balance-perturbation protocol.

Task conditions	Focus of analysis	Perturbations and numbers of trials[a]	
		Analysed trials	Additional trials
Trial block #1: stance, arm motion restricted, instructed to react naturally but minimise number of steps if need to step	Stepping evoked by AP perturbation	5 backward translations[b] 3 forward translations 5 forward cable-pulls[c] 3 backward cable-pulls Total = 16 trials	2 ML translations (L,R) 4 translations, 2nd waveform (F,B,L,R)[d] 2 ML cable-pulls (L,R) Total = 8 trials
Trial block #2: walking in-place[e], arm motion restricted, instructed to react naturally but minimise number of steps if need to step	Stepping evoked by ML perturbation	5 leftward translations 3 rightward translations 5 rightward cable-pulls 3 leftward cable-pulls Total = 16 trials	2 AP translations (F,B) 4 translations, 2nd waveform (F,B,L,R)[d] 2 AP cable-pulls (F,B) Total = 8 trials
Trial block #3: stance, foot motion restricted, instructed to recover balance by grasping handrail (at right of subject)	Grasping evoked by backward perturbation	5 forward translations 5 backward cable-pulls Total = 10 trials	4 backward translations 4 forward cable-pulls Total = 8 trials

[a] During each trial block, the listed surface-translation (ST) and cable-pull (CP) perturbations were delivered in an unpredictable randomized sequence, in the directions indicated (F = forward, B = backward, L = left, R = right; AP = antero-posterior, and ML = medio-lateral). The "additional trials" were included solely for the purpose of increasing unpredictability and were not analysed. Given 40 subjects, the specified protocol yields the following numbers of trials for analysis: 640 AP-step trials (40×16) in trial-block #1, 640 ML-step trials (40×16) in trial-block #2, and 400 grasp trials (40×10) in trial-block #3. However, the actual numbers of trials performed were reduced to 622 in trial-block #2 and 364 in trial-block #3, due to a small number of subjects who wished to terminate the session early.

[b] The ST perturbations comprised a 300 ms acceleration pulse followed immediately by a 300 ms deceleration pulse (Fig. 2A). Each pulse was approximately "square", with an amplitude of 2.0 m/s^2 for forward translations (evoking backward falling motion) and 3.0 m/s^2 for other translation directions (backward, left, and right). The displacement and peak velocity were 0.18 m and 0.6 m/s for forward translations; 0.27 m and 0.9 m/s for other directions.

[c] The CP perturbations were applied by dropping a weight equal to 20% of body weight. The drop height was 40 cm for stepping trials and 30 cm for grasping trials.

[d] The "2nd waveform" STs were included to deter subjects from learning to use the platform deceleration to aid in recovering balance (McIlroy and Maki, 1994). This waveform comprised a 200 ms acceleration pulse, a 400 ms constant-velocity interval, and a 200 ms deceleration pulse. The acceleration of this second waveform ranged from 1.35 to 2.25 m/s^2.

[e] Perturbations for the walking-in-place trials were timed to occur at foot-lift of the foot contralateral to the fall direction evoked by the perturbation (e.g. right foot-lift for a leftward fall) in order to increase the probability of observing a collision between the step foot and the stance leg (Maki et al., 2000). The perturbation was delivered after a random number of steps (3–8) and was triggered when the stance-leg force-plate loading exceeded 90% of body weight.

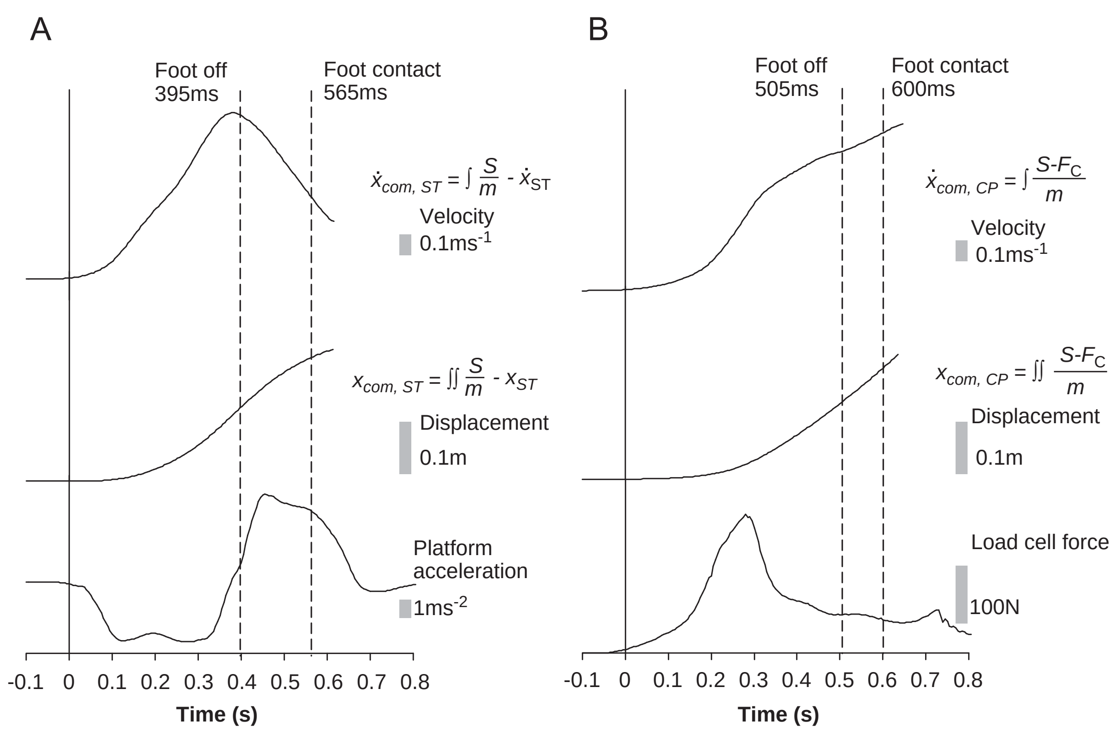

Fig. 2. Centre-of-mass (COM) displacement and velocity for typical trials (young–adult subject), for the two perturbation methods: (A) surface-translation (ST) and (B) cable-pull (CP). The plots at the bottom of each panel show the time history of the surface acceleration and CP force. The equations that were integrated to determine the COM motion are included on the graphs. In these equations, $x_{COM,ST}$ and $x_{COM,CP}$ are the antero-posterior (AP) COM displacements (relative to the motion platform) for ST and CP perturbations, S is the total AP shear force recorded from the force plates, m is the body mass, x_{ST} is the AP motion-platform displacement recorded during ST perturbations, and F_c is the AP cable force recorded during CP perturbations. Numerical integration of the equations began at time zero, which corresponds to perturbation onset (i.e. platform acceleration >0.1 m/s^2 or cable force >5 N). Previous studies using this same equipment have shown that this method of estimating COM velocity and displacement is reliable provided that the duration of the integration is limited to ~800 ms; however, significant propagation of errors due to signal drift can occur when integrating force-plate signals over longer time periods (Maki and McIlroy, 1999b; McIlroy and Maki, 1999). To avoid this problem, we terminated the integrations at time of foot-contact (which always occurred within the 800 ms limit).

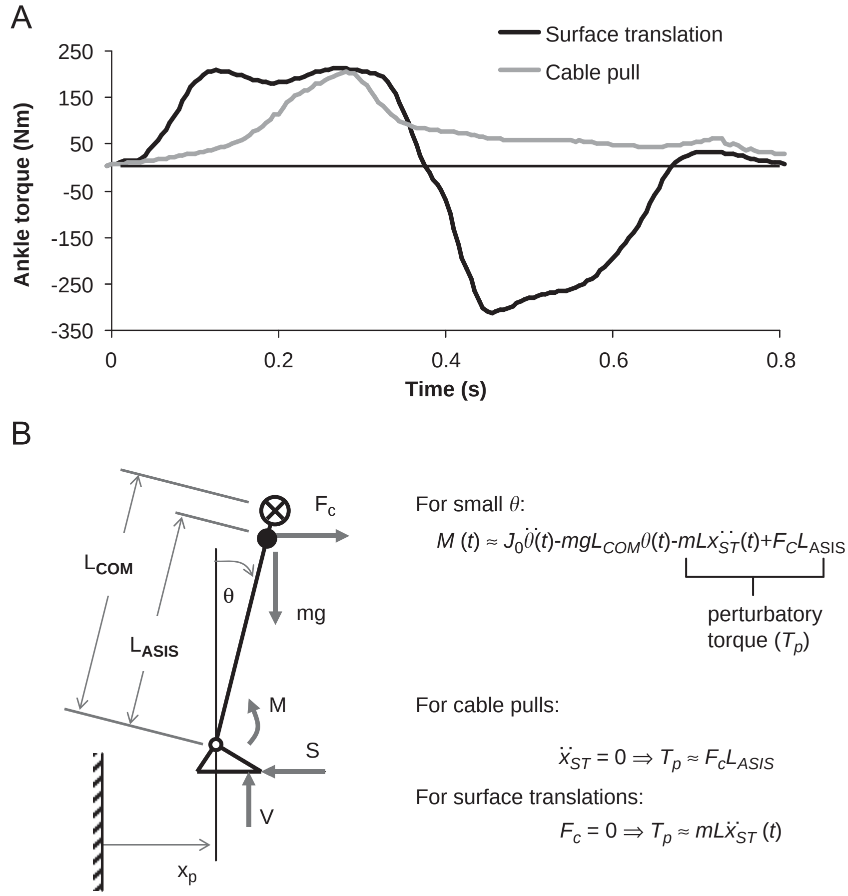

Fig. 3. Comparison of the biomechanical effects of the two perturbation methods. Panel (A) compares the perturbatory ankle-torque resulting from surface-translation (ST) and cable-pull (CP) perturbations, for typical trials in a young–adult subject (body mass 70.5 kg, height 1.75 m). Panel (B) shows the simple inverted-pendulum model and equations of motion used to calculate these torques (T_p), where M is the muscle-generated ankle moment, J_0 is the rotational inertia about the ankle, θ is the ankle-joint angle, m is the body mass, g is gravitational acceleration, L_{COM} is the distance between the COM and the ankle joint, L_{ASIS} is the distance between the CP cable (attached to the pelvic belt worn at the level of the anterior–superior iliac spine) and the ankle joint, x_{ST} is the displacement of the motion platform recorded during ST perturbations, and F_c is the cable force recorded during CP perturbations. The equations of motion are based on modelling work by Maki (1987), and the anthropometric and inertial parameters were estimated using data tabulated by Drillis and Contini (1966). Note that the peak value of T_p was similar for the ST- and CP-trials; however, there was a much more rapid rise in T_p in the ST-trial. Note also the reversal in T_p that occurs during the last half of the ST-trial; this reflects the stabilising effect of the platform deceleration (McIlroy and Maki, 1994).

Table 3
Effect of perturbation method on centre-of-mass (COM) displacement and velocity.

	COM parameter	Point in time	Young adults			Older adults		
			Mean (SD)			Mean (SD)		
			ST	CP		ST	CP	
Forward "falls"[a]	Forward COM displacement (cm)	100 ms after PO[b]	0.89 (0.88)	0.21 (0.09)	**	1.00 (1.32)	0.29 (0.24)	**
		Foot-off	15.7 (6.8)	18.1 (5.6)	**	11.9 (3.8)	14.1 (4.3)	**
		Foot-contact	24.8 (6.5)	31.9 (7.4)	**	21.9 (5.6)	25.0 (5.9)	**
	Forward COM velocity (cm/s)	100 ms after PO	9.6 (1.4)	3.5 (1.0)	**	9.6 (1.4)	4.0 (1.9)	**
		Foot-off	69.5 (17)	72.4 (10.0)		76.9 (5.2)	65.5 (7.9)	**
		Foot-contact	39.7 (13.9)	94.3 (15.7)	**	48.8 (14.6)	84.9 (11.9)	**
Backward "falls"[a]	Backward COM displacement (cm)	100 ms after PO	0.92 (0.98)	0.25 (0.17)	**	0.87 (0.72)	0.39 (0.51)	**
		Foot-off	11.3 (5.5)	13.6 (3.1)	**	9.7 (4.5)	11.6 (4.6)	**
		Foot-contact	18.3 (4.4)	26.6 (5.5)	**	15.7 (4.3)	21.0 (6.7)	**
	Backward COM velocity (cm/s)	100 ms after PO	11.8 (2.0)	3.6 (1.6)	**	11.4 (1.5)	4.5 (3.0)	**
		Foot-off	58.0 (9.0)	68.1 (8.7)	*	60.2 (8.4)	58.9 (11.4)	
		Foot-contact	25.5 (20.9)	91.0 (15.1)	**	37.4 (15.4)	75.7 (14.6)	**

Significant effect (within the specified age group) due to perturbation-type:

* $p<0.05$;

** $p<0.001$.

[a] Forward "falls" were induced by backward surface translations (STs; 200 trials) or forward cable-pulls (CPs; 200 trials). Backward "falls" were induced by forward STs (120 trials) or backward CPs (120 trials).

[b] PO = perturbation onset.

Table 4
Summary of age effects for surface-translation and cable-pull perturbations.

	Number of trials analysed[a]	Surface-translation (ST) trials			Cable-pull (CP) trials			Age-method interaction (p-value)[c]
		Mean (SD)			Mean (SD)			
		YA	OA	Effect size[b]	YA	OA	Effect size[b]	
Trial-block #1: stepping evoked by AP perturbation of stance								
"Extra" lateral steps (% trials)	640	3.8 (19.1)	19.2 (39.5)	−0.43**	0.0 (0.0)	1.3 (11.2)	−0.13	**0.005**
Earliest TA latency (ms)[d]	224	98 (23)	169 (64)	−2.45***	137 (24)	166 (62)	−0.47	**0.0001**
Earliest MG latency (ms)[d]	394	127 (26)	176 (56)	−1.58***	153 (30)	199 (68)	−0.70	**0.053**
Foot-off time (ms)	639	387 (85)	353 (41)	0.61**	521 (80)	473 (71)	0.63*	0.93
Foot-contact time (ms)	639	545 (83)	492 (57)	0.78**	682 (79)	614 (77)	0.83	>*0.99*
Swing duration (ms)	639	158 (36)	140 (42)	0.44	161 (39)	140 (40)	0.51	0.58
Presence of an APA (% trials)	639	73.8 (44.3)	53.3 (50)	0.41**	86.1 (34.8)	71.7 (45.2)	0.33	0.51
Step length (cm)	628	33.5 (9.4)	29.0 (12)	0.39	37.1 (9.8)	28 (10.7)	0.81***	**0.01**
Step width (cm)	628	2.8 (4.2)	1.7 (3.1)	0.33	0.8 (3.5)	1.2 (2.9)	−0.15	**0.045**
Trial-block #2: stepping evoked by ML perturbation while walking in place								
Foot collisions (% trials)	619	26.3 (44.3)	55.6 (49.8)	−0.59*	1.3 (11.3)	5.7 (23.2)	−0.09	0.21
Crossover steps (% trials)	616	41.3 (49.5)	30.7 (46.2)	0.50	56.4 (49.9)	32.2 (46.8)	0.50	0.39
Trial-block #3: grasping evoked by backward perturbation of stance								
Right bicep latency (ms)	341	116 (30)	147 (31)	−0.95**	152 (53)	177 (62)	−0.42	0.35
Right deltoid latency (ms)	349	115 (26)	139 (30)	−0.79**	152 (50)	173 (53)	−0.40	0.43
Handrail contact time (ms)	349	494 (61)	546 (88)	−0.62*	602 (94)	654 (154)	−0.37	0.43
Pooled stepping trials (trial-blocks #1 and #2)								
Multi-step reactions (% trials)	1254	25 (43.4)	68.3 (46.6)	−0.87***	26.6 (44.3)	50.0 (50.1)	−0.47**	**0.028**
Arm reactions (% trials)	1254	1.3 (11.1)	16.8 (37.4)	−0.46**	0.0 (0.0)	1.5 (12.2)	−0.14	**0.017**
Pooled stepping and grasping trials (trial-blocks #1, #2, and #3)								
Harness-assist (% trials)[e]	1626	0.0 (0.0)	0.2 (4.1)	−0.05	0.0 (0.0)	1.0 (9.9)	−0.12	0.31

[a] Discrepancies between number of trials actually analysed and trial numbers specified in Table 2 were due to technical problems.

[b] Effect size is the difference between the young (YA) and older-adult (OA) means (YA minus OA), divided by the pooled standard deviation. Results from one-way ANOVA indicating significant age-group effects: $^{*}p<0.05$, $^{**}p<0.001$, and $^{***}p<0.0001$.

[c] Results from the two-way ANOVA indicating statistical significance of the age-group by perturbation-method interaction. Significant and near-significant interactions are given in bold.

[d] Earliest tibialis anterior (TA) latency analysed for "backward fall" trials (i.e. backward CPs and forward STs); medial gastrocnemius (MG) latency analysed for "forward fall" trials (i.e. forward CPs and backward STs).

[e] A harness-assisted recovery was deemed to occur if the safety harness loading exceeded 20% of body weight.

peak torque amplitude was similar (Fig. 3). Later phases of the perturbations also differed. For STs, the initial perturbatory torque resulting from the support surface acceleration was followed by a perturbation in the opposite direction due to surface deceleration. Conversely, each CP provided a unidirectional perturbation.

Analysis of COM motion (AP-perturbation trials) confirmed the more rapid rise in destabilising effect of STs; COM displacement and velocity 100 ms after perturbation onset (before the postural reaction) were higher for STs than CPs (p-values <0.0001; Table 3). Conversely, COM displacement at foot-off and foot-contact time was larger in CP-trials (p-values <0.0001). CP-trials also showed higher COM velocity at foot-contact (p-values <0.0001); however, at foot-off, this trend was only seen in YA "backward-fall" trials ($p=0.026$), and CPs showed lower COM velocity in OA "forward-fall" trials ($p<0.0001$).

Age-related differences in change-in-support responses were commonly observed, but were more pronounced in ST-trials (Table 4; Fig. 4). A small number of variables failed to show any significant age-effects, in either ST- or CP-trials (frequency of harness-assisted recoveries, swing duration and step width in AP-step trials, frequency of crossover steps in ML-step trials). With one exception (AP-step length), STs revealed statistically significant ($p<0.05$) age-effects for all remaining variables. CP-trials invariably demonstrated the same age-related trends in these variables; however, the age-effect sizes were smaller and often failed to attain statistical significance. AP-step length was the only variable to show a stronger, statistically significant age-effect in CP-trials (the age-effect in ST-trials was in the same direction but smaller; Fig. 4E).

STs appeared more likely than CPs to cause problems for older adults, evidenced by the higher frequency of: (1) arm reactions (despite constraints on arm movement) during stepping responses (17% ST, 2% CP); (2) "extra" lateral steps in AP-perturbation trials (19% ST, 1% CP); (3) collisions between the swing and stance limbs in ML-perturbation trials (56% ST, 6% CP). "Floor effects" in these CP data likely contributed to small age-effect sizes (0.09–0.14) and the failure to attain statistical significance.

The dependence of age-effect on perturbation method was confirmed by two-way ANOVA. These analyses showed a significant age-by-method interaction ($p<0.05$) for: frequency of multi-step reactions; frequency of arm reactions during stepping trials; tibialis-anterior latency, step length, step width, and frequency of "extra" lateral steps in AP-perturbation trials. In all

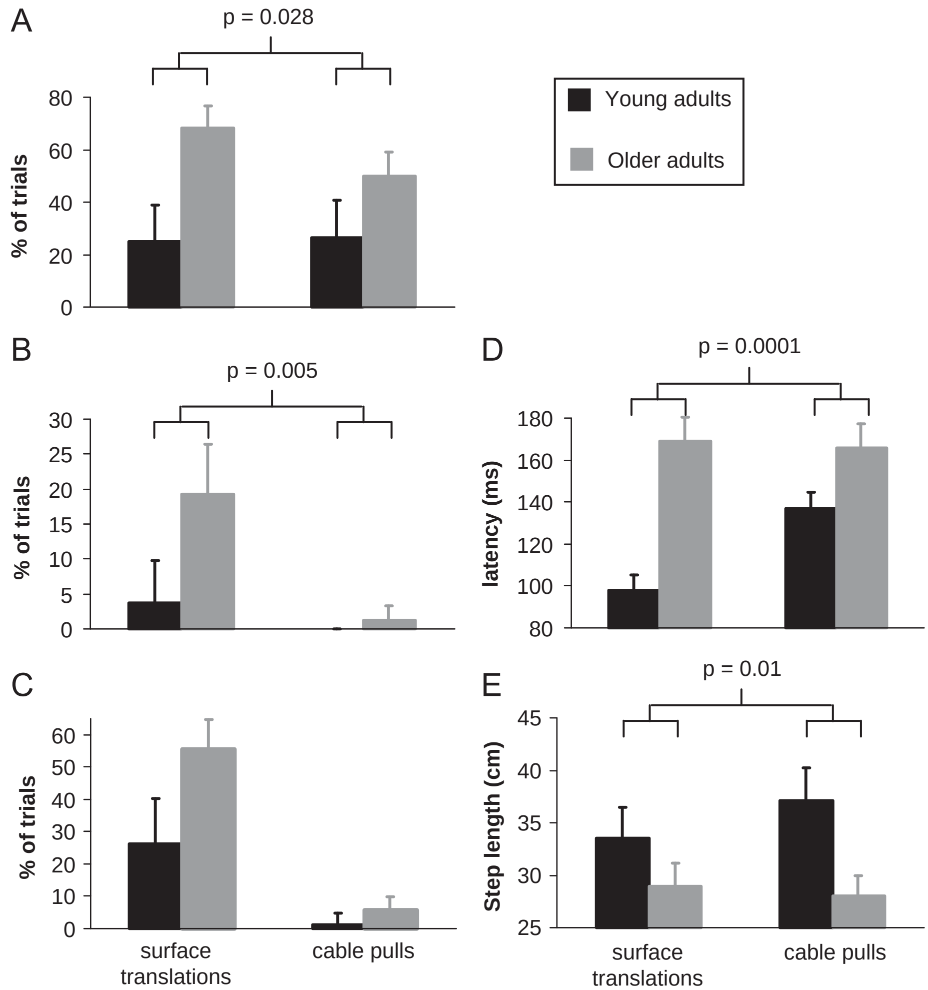

Fig. 4. Example results: (A) percentage of trials in which multi-step reactions were observed; (B) percentage of AP-perturbation trials in which an "extra" lateral step (subsequent to the initial forward or backward step) was observed; (C) percentage of ML-perturbation trials in which a collision between the swing and stance limbs was observed; (D) tibialis-anterior latency ("forward fall" trials); (E) AP step-length (AP-perturbation trials). In each graph, means and standard-error bars are shown for the young and older adults, for the surface-translation (ST) perturbations and the cable-pull (CP) perturbations. For the frequency (percentage-of-trial) variables, a percentage score was determined for each perturbation method within each individual subject, and these scores were then used in the subsequent analyses. The interaction *p*-values are indicated in cases where the two-way ANOVA revealed a significant interaction between age-group and perturbatio method. Note that certain age-related problems were common in the ST-trials but occurred rarely, if at all, in the CP-trials (panels B and C). The remaining plots appear to exhibit age-related effects for both perturbation methods. The age-effect was more pronounced in ST-trials in panels A and D, as was the case in all but one of the analyses (see Table 4). Panel E depicts the one exception, where the age-effect was more pronounced in the CP-trials.

but one case (AP-step length), the interaction indicated a stronger effect in the ST-trials (Fig. 4).

4. Discussion

Although age-effects were almost always more pronounced in ST-trials, the direction of the age-effect (i.e. whether the variable increased or decreased in OA) was invariably the same for both perturbation methods, as hypothesised. The similar nature of age-effects across two distinctly different perturbation methods suggests that the specific nature of the mechanical stimulus and associated sensory drive were not critical factors.

The more pronounced age-related differences observed in ST-trials suggest that the STs were more destabilising than the CPs. Supporting this, our inverted-pendulum model predicted a more rapid rise in perturbatory ankle-torque in ST-trials. Although this was a very simple model, COM data were consistent with the model predictions, showing larger COM displacement/velocity 100 ms after perturbation onset in ST-trials. These COM data primarily reflect the destabilising effect of the perturbation, as the earliest component of the perturbation-evoked reaction occurred ~100 ms or more after perturbation onset.

Interpretation of COM motion at later points is more complicated as this also reflects the influence of the perturbation-evoked reaction. Although there were mixed results at foot-off, COM displacement and velocity at foot-contact were consistently higher for CPs. This could reflect the stabilizing influence of platform deceleration in ST-trials (McIlroy and Maki, 1994). The platform deceleration pulse began 300 ms after perturbation onset, and hence was likely too late to substantively affect COM motion prior to foot-off (~350–400 ms after perturbation onset) but could have reduced COM motion prior to foot-contact (~500–550 ms after perturbation onset). This may, in turn, have reduced the step length required to recover equilibrium, thereby decreasing the likelihood of exposing age-related limitations in ability to rapidly generate large steps (Hsiao-Wecksler and Robinovitch, 2007). Such effects could explain the finding that age-related differences in AP-step length were less pronounced in ST-trials; however, this cannot be determined conclusively from the present data.

The modelling and COM results suggest that perturbation time-course (waveform) was a more important factor than the nature of mechanical and sensory stimuli. Thus, for example, the slower rise in perturbatory ankle-torque generated by CPs allowed OA more time to plan and execute change-in-support reactions, whereas the more urgent need to respond rapidly to ST perturbations would be more likely expose limitations arising from age-related speed-of-processing decrements (Schulz et al., 2006). This is supported by the fact that foot-off, foot-contact, grasping-reaction latency, and rail-contact were all delayed in CP-trials (Table 4). The more rapid onset of STs, compared to CPs, may have also contributed to the higher frequency of limb collisions during ML-step trials. In these trials, subjects walked in-place with the perturbation triggered to occur shortly after one foot was lifted. The very rapid initial ML support surface movement caused the stance leg to move rapidly toward the lifted foot, making it difficult to avoid a collision between the limbs.

The unpredictable trial-to-trial variation in perturbation method (CP or ST) is unique to this study. Potentially, this higher level of unpredictability could be responsible for some discrepancies between this and previous studies. For example, previous studies have shown very small age-related delays (e.g. <10 ms, Woollacott et al., 1986) in onset of early "automatic postural responses" in ankle musculature, whereas the present study showed much larger age-effects ($\sim$40–50 ms). Conceivably, the higher degree of unpredictability may have impeded the ability of OA to detect onset of instability, compared with other studies.

To facilitate comparison with previous studies, the present findings are included in Table 1. Our findings agree with previous ST studies, with one exception: the present analysis of foot-off times showed more rapid initiation of AP steps in OA, whereas the previous work indicated no significant age-effect for foot-off time (McIlroy and Maki, 1996). The YA foot-off times were quite similar in both studies, whereas our OA initiated more rapid responses than the previously studied OA. The discrepancy could be due to differences in perturbation unpredictability (greater in the present study) and/or differences in OA cohort (history of instability/falls, no such history in the previous study).

For CPs, the present results are supported by previous studies in some cases but not in others. There was mixed support for our findings that AP-step foot-off times were faster in OA and that swing duration was unaffected by age; however, there was agreement with our finding that OA took shorter steps. For ML-step reactions, a previous CP study found that OA favoured crossover steps and that crossovers were associated with a high rate of limb collisions (Mille et al., 2005), whereas we found no significant age-effect on frequency of crossover steps and limb collisions occurred rarely in our CP-trials. The ML-step discrepancies could be due to the fact that our ML perturbations were delivered during in-place walking (rather than bipedal stance). It is more difficult to pinpoint the cause of discrepancies in AP-step findings. Possible explanations include differences in: cohort characteristics, perturbation unpredictability, perturbation waveform, instructions, measurement/analysis methods, and/or statistical power.

Regarding grasping reactions, only one previous study has examined age-related differences; that study used AP ST-perturbations (Maki et al., 2001). The present ST-results agree with these findings, indicating age-related delays in initiation and completion of grasping reactions. The present CP data show trends in this same direction; however, these findings were not statistically significant. The present results also agree with previous ST step-reaction studies showing more frequent arm reactions in OA (Maki et al., 2001), but showed that such arm reactions were very infrequent in CP-trials. This could be due to the less destabilising nature of the CP waveform, but could also possibly reflect a reluctance, in ST-trials, to rely on stepping to recover balance when standing on an "unreliable" moving surface.

Although we did not include release-from-lean perturbations in this initial study, it is evident, from Table 1, that previous RFL studies have yielded substantially different findings. The predictability of the RFL perturbations is likely to be a major contributing factor. Results may also differ because the pre-perturbation ankle dorsiflexion required in RFL tests (up to 40°) may limit the degree to which ankle rotation can occur during the balance-recovery reaction. Furthermore, the forward COM displacement associated with the initial lean posture necessitates either a more rapid step or a larger step distance, in comparison to step reactions evoked when subjects are not leaning (Maki and McIlroy, 1999a). The extreme level of challenge associated with large lean angles may thus explain why RFL studies (Thelen et al., 1997; Wojcik et al., 1999) have been successful in exposing age-related limits on step-initiation speed (not seen in CP and ST studies). Conversely, the age-related increase in step length seen in RFL studies may represent preplanned efforts to compensate for delayed step initiation. Future studies could potentially disentangle confounding effects of perturbation predictability by incorporating more unpredictability into the RFL protocol (e.g. by mounting the RFL apparatus on a motion platform, so that ST perturbations can also be delivered).

The present study is the first to directly compare age-related differences in change-in-support reactions evoked using different perturbation methods. The results are directly applicable to studies using the specific perturbation methods tested (weight-drop CPs and motor-driven STs), but cannot definitively isolate effects due to differences in perturbation waveform versus the nature of the perturbation *per se*. Future studies may take advantage of more sophisticated electromechanical actuators (Pidcoe and Rogers, 1998) to create CP waveforms that more closely match the ST waveforms, allowing the influences due to perturbation type and waveform to be de-coupled.

Another potential limitation of the present study pertains to the single-joint inverted-pendulum model. This model is a simplification of the multi-segmental motion that can occur during balance-recovery reactions, but has nonetheless been shown to predict stepping behaviour evoked by both CPs and STs (Pai et al., 1998, 2000). We would also emphasize that this model was intended only to provide a means of identifying major differences in perturbation time-course, and was not intended to provide precise predictions of joint kinematics and kinetics. A third limitation is that the comparison of COM motion was restricted to AP-perturbations. We elected not to analyze COM motion evoked by ML-perturbation, in this study, because the interpretation of these data is complicated by variation in: (1) pattern of stepping (crossover versus side-step); (2) occurrence of collisions between the swing and stance limbs; (3) ML COM velocity at time of perturbation onset (due to variability in the "walk-in-place" pattern). The degree to which the two perturbation methods induce different patterns of ML COM motion remains to be addressed in future studies.

In conclusion, the present results indicate that age-related differences in the control of perturbation-evoked change-in-support reactions were not critically dependent on the nature or point-of-application of the perturbatory force, as both of the tested perturbation methods consistently yielded the same direction of age-effect. The strength of the effect, however, appeared to be highly dependent on the perturbation waveform, i.e. features such as rise-time and phase-reversal (e.g. due to support surface deceleration). For the specific ST and CP waveforms tested here, the ST perturbations were more effective in revealing age-related differences.

Conflict of interest statement

The authors declare that they have no conflicts of interest.

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