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Mechanisms underlying age-related differences in ability to recover balance with the ankle strategy

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Abstract

Falls cause substantial death and morbidity in the elderly. An important prerequisite to the development of fall prevention interventions is improved understanding of the biomechanical and neuromuscular variables that govern ability to recover balance. In the present study, we examined the relative importance of strength versus speed of response variables in explaining differences in balance recovery performance with the ankle strategy between young and elderly women. Twenty-five young (19–36 years) and 25 community-dwelling elderly women (66–90 years) participated in balance recovery experiments. Subjects were supported in an inclined standing position by a horizontal tether and instructed to recover an upright vertical standing position by contracting their ankle muscles. The maximum initial lean angle from which they could recover balance without release of the tether (which depends primarily on strength) was 19.6% smaller for elderly than young. The maximum initial lean angle from which they could recover balance after the tether was suddenly released (which depends on both strength and speed of response) was 36.1% smaller for elderly than young. Moreover, between-group differences in performance were related to both strength and speed of response. Peak ankle torque was 7.7% smaller in elderly than young during dynamic recovery trials; reaction time was 27% slower in elderly, due to a lengthened muscle response latency, and rate of ankle torque generation was 15.6% slower in elderly. These results suggest that differences in ability to recover balance between young and elderly women associate with variables related to strength and speed of response, and that exercise-based fall prevention programs should include balance and agility training, in addition to strength training.

Keywords: Accidental fall; Musculoskeletal equilibrium; Reaction time; Strength; Biomechanics

1. Introduction

Falls are the leading cause of injury and injury-related deaths in the elderly [1]. While fall risk depends on a variety of sensory, motor, cognitive, psychosocial, and environmental variables [2,3], it depends ultimately on one's frequency of imbalance episodes, and one's ability to recover balance after such episodes [4]. Therefore, fall prevention programs should target each of these areas.

An important prerequisite to the development of fall prevention programs is an improved understanding of the biomechanical and neuromuscular variables that govern ability to recover balance. Both strength [5] and speed of response to loss of balance [6–8] are known to decline with age, and these changes lead to increased fall risk [2,3,9]. However, the relative importance of strength versus speed of response variables in explaining age-related changes in ability to recover balance is unknown. It is important to identify the declines with age in the relative contribution to balance recovery of each of these variables, since fundamentally different types of interventions (e.g., exercise programs) may be required to target each of these variables.

Previous studies suggest that strength variables, such as peak attainable torque, and speed of response variables, such as reaction time and rate of torque development, are important in determining young subjects' ability to recover balance [4,10,11]. In this study, we hypothesized that ability to recover balance is smaller in elderly women with a history of falls than in healthy young women, and these differences

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in recovery ability reflect the combined effect of differences in strength variables and speed of response variables. To address these hypotheses, we conducted balance recovery experiments with young and elderly women and measured the maximum forward lean angle where subjects could recover a stable upright stance by contracting their ankle muscles (ankle strategy). We examined this index of balance recovery ability when subjects self-initiated their recovery versus recovered balance after being suddenly released from a forward leaning position. The former parameter, which was termed ''maximum static recovery angle'', should depend primarily on factors related to muscle strength, while the latter parameter, which was termed ''maximum dynamic recovery angle'', should depend on factors related to both muscle strength and speed of response. The percent difference between young and elderly in the maximum static recovery angle was therefore used to determine the effect of declines in strength, while the percent difference between young and elderly in the maximum dynamic recovery angle was used to determine the effect of declines in strength and speed of response. In addition, we also examined performance variables from the recovery trials, including peak ankle torque, reaction time, and rates of ankle torque generation and decline, and we associated these variables with the maximum dynamic recovery angle to further determine the effect of declines in strength and speed of response.

2. Methods

2.1. Subjects

Twenty-five healthy young women of mean age 25 ± 4 (S.D.) years (range 19–36 years), and 25 healthy elderly women of mean age 78 ± 7 years (range 66–90 years) participated in this study. Young subjects were recruited through notices at Simon Fraser University, while elderly subjects were recruited from an education-based fall prevention program that operated between autumn 2002 and winter 2003 in local seniors' centres. To be eligible for the study, elderly subjects must have reported at least one fall, as described in the FICSIT definition [12], either in the 12 months prior to their entry into the fall prevention program, as recorded on their intake questionnaire, or since their entry into the program, as recorded on their monthly fall passports. These subjects represented a relatively high-risk group for future falls.

Subjects were initially interviewed in-person and individuals were excluded if they had conditions that would prevent them from performing the experiments, or possessed known risk factors for falls that would represent confounding variables in the analysis, since our intent was to determine factors that affect risk for unexplained falls and balance impairments. These included terminal illness; inability to stand unassisted for 10 min; physical deformity of the lower limbs or spinal column which affect stability (including leg lengths that differed by more than 10 cm, foot lengths that differed by more than 5 cm, foot deformities, tendon abnormalities, and scoliosis); blindness; Parkinson's disease, stroke; peripheral neuropathy or plantar foot ulcers; profound and recurring episodes of vertigo, dizziness, or loss of consciousness in the past 3 months; use of psychotropic medications (hypnotics, anxiolytics, antidepressants, and antipsychotics) [13]. Subjects were excluded after an on-site evaluation if they had (1) moderate to severe dementia (Folstein Mini-Mental State Examination score less than 21) [14], (2) major uncorrected deficits in visual acuity (Snellen score worse than 20/15 at 5 ft.), (3) evidence of profound vestibular deficits (gross instability when standing on foam with eyes closed) or (4) evidence of profound somatosensory deficits (measured by big toe position sense and monofilament to the dorsum of the foot). Each subject was paid CDN$ 10/h for her participation. Each subject provided informed written consent and the experiment was approved by the University Research Ethics Board at Simon Fraser University.

2.2. Ancillary measurements

Elderly subjects visited the laboratory for two sessions (usually no more than 1 week, and at the most, 2 weeks apart), while young subjects visited once. During the first session for elderly subjects, and the first half (approximately 1.5 h) of the session for young subjects, we acquired ancillary measures to characterize the samples. We measured get-up and go time (6 m) [15] (young = 9.8 ± 1.5 s; elderly = 14.2 ± 2.7 s), functional reach [16] (young = 38.1 ± 4.4 cm; elderly = 27.4 ± 5.9 cm), Folstein Mini-Mental State (scored 0–30) [14] (young = 29 ± 1; elderly = 27 ± 2), and Activities-specific Balance Confidence (scored 0–100%) [17] (young = 98 ± 3%; elderly = 79 ± 16%).

2.3. Experimental procedures

The experimental methods are similar to those reported by Robinovitch et al. [4]. We conducted balance recovery experimental trials in which we measured the maximum recovery angle from which subjects were able to recover balance (a stable upright vertical stance) primarily by contracting the muscles spanning the ankle joint, a technique called the ''ankle strategy'' [18,19]. To conduct the experimental trials, the subject stood on a force plate with her feet a comfortable width apart and arms crossed against her chest. We then inclined the subject into a stationary forward leaning posture via a horizontal tether that attached at one end to an electromagnetic brake (Warner Electric model PB500, South Beloit, IL), and at the other end to a chest harness worn by the subject (Fig. 1A). As a safety precaution, we also attached an overhead fall restraint tether to the chest harness. We instructed the subject to return to a vertical standing position by contracting the muscles spanning the ankles, keeping the knees and hips fully extended, the heels were allowed to leave the ground. Practice trials were provided until the subject understood the requirements of the ankle strategy and was comfortable leaning into the tether before release. This was typically achieved in three to six practice trials.

To determine the effect on recovery ability of the magnitude versus speed of torque development, the maximum recovery angle was measured under ''static'' and ''dynamic'' conditions. In static trials, the subject attempted to simply return to a standing position when the tether was not released (Fig. 2A). Following a ''ready'' cue from the subject, the investigator provided an auditory tone, signaling the subject to begin the recovery process. In dynamic trials, the subject attempted to recover a stable upright position after the tether was suddenly released, which caused a near-step increase in the resultant torque acting to rotate the body downward (tether release time ~15 ms) (Fig. 2B). To increase the unexpectedness of the perturbation, the investigator inserted a random time delay of 1–10 s between receiving the ''ready'' cue from the subject and the time of release. In both static and dynamic conditions, the first trials involved small lean angles of ~2°, where the subject could recover easily. We then iteratively adjusted the length of the tether and the corresponding lean angle until we identified the maximum recovery angle (with a resolution of 7 mm in tether length and ~0.3° in lean angle), beyond which the subject could no longer recover balance in at least three of five repeated trials. The order of presentation of the two different experimental conditions was counterbalanced across subjects to minimize order or learning effects. Before each trial, the subject was instructed to maintain her gaze forward and at eye level on a black × attached to the wall approximately 10 ft. in front of her.

Subjects were barefoot during the trials and wore tight-fitting shorts and shirt. To offset the possibility of subject fatigue, rest breaks of approximately 30 s duration were provided between trials, and 5 min sitting breaks were provided between the different conditions. In static trials, we monitored tether force to ensure the subject did not use the tether to aid her recovery performance. During dynamic

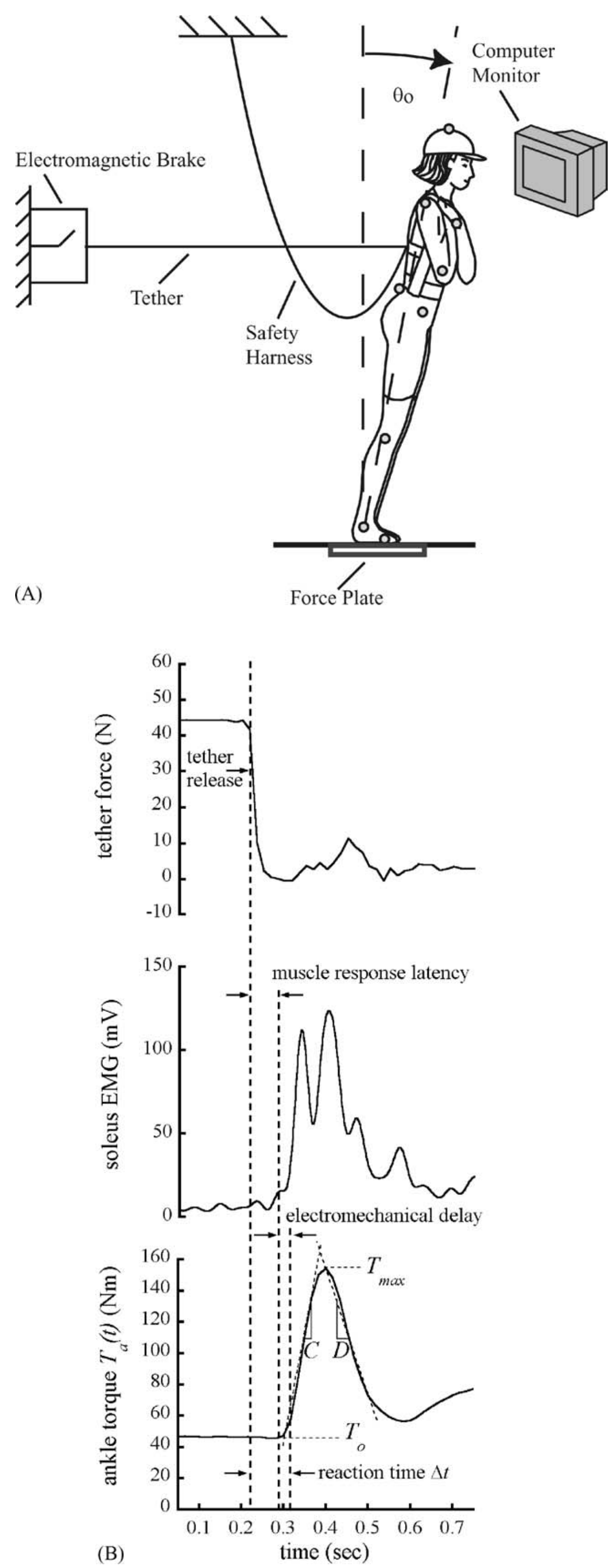

Fig. 1. (A) Balance recovery experiment. Subjects were held by a tether in an initially inclined position and instructed to recover a stable upright position by contracting their ankle muscles. Their ''maximum static recovery angle'' was the maximum initial forward lean angle (θ_0) where they could accomplish this task when the tether was not released. Their ''maximum dynamic recovery angle'' was the maximum θ_0 where they could recover balance after the tether was suddenly released. (B) Ankle torque profile characteristics and measured reaction time components. We detected tether release (shown by the earliest vertical dashed line) by a sharp decline in tether force. Before tether release, subjects adjusted their ankle torque to match the average value during quiet standing (T_0) (lower panel). Following release, there was a reaction time (Δt) before the onset of increased ankle torque generation. Ankle torque was generated at a rate C and reached a peak, T_{max}, before declining at a rate D. Total reaction time (Δt) was composed of muscle response latency and electromechanical delay. Onset of increased EMG activity in the soleus is shown by the second vertical dashed line, and onset of increased ankle torque production is shown by the third vertical dashed line. We regarded muscle response latency, EMD, Δt, C, and D as speed of response variables, and T_{max} as a strength variable.

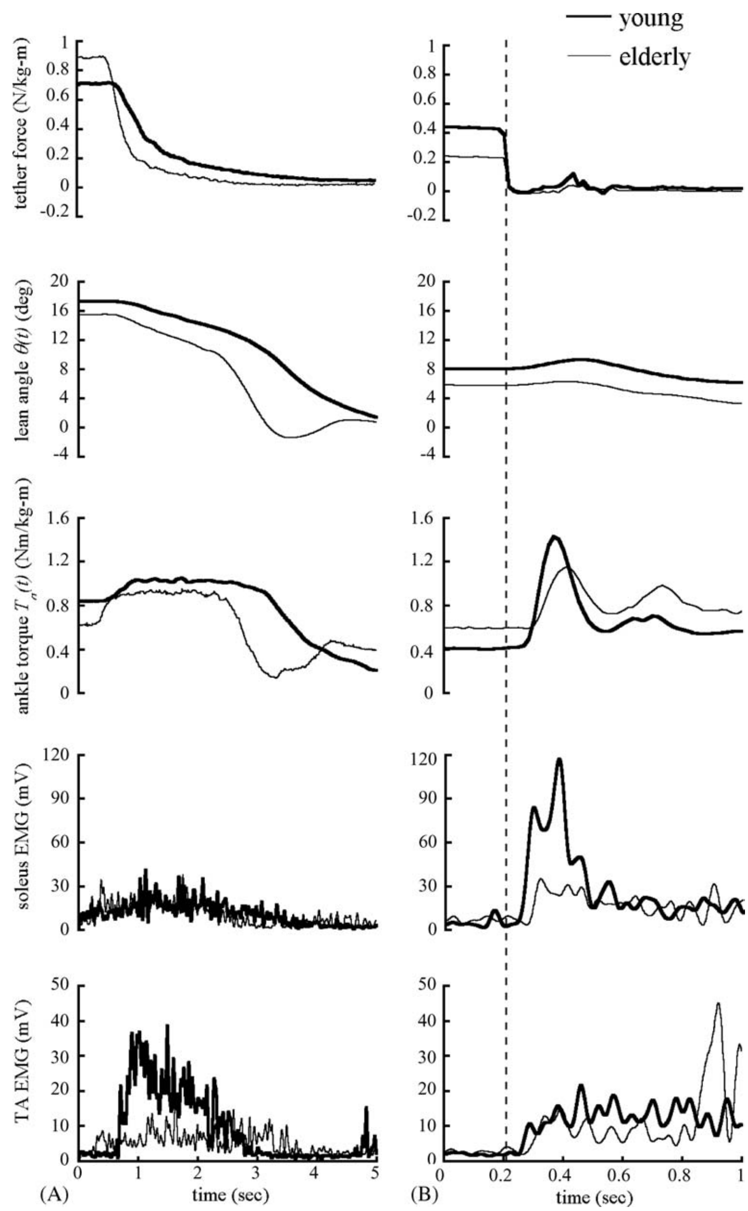

Fig. 2. Typical variations in kinematic and kinetic parameters for a 22-year-old young subject (heavy traces) and a 66-year-old elderly subject (light traces) in (A) static recovery trial and (B) dynamic recovery trial. In the static trial, each subject self-initiated her recovery into a stable upright position by increasing her ankle torque. This caused the body lean angle to decrease and the tether force to decline. The young subject generated a greater peak ankle torque than the elderly subject, and a greater maximum recovery angle. In the dynamic trial, tether release (indicated by the dashed line) was followed by a sharp decline in tether force. After release, the body rotated downward (lean angle increased), and after a time delay of 88 ms for the young subject and 107 ms for the elderly subject, ankle torque began to increase and halt downward rotation of the body, which allowed for return to upright posture. The soleus muscle response latency was longer for the elderly subject, and consequently, the onset of ankle torque generation was later. Also, the rate of ankle torque generation and the peak ankle torque were smaller for the elderly subject. Note that while peak ankle torque was greater in the dynamic trial than the static trial for both subjects, their maximum recovery angles were smaller in the dynamic trials.

trials, we controlled the magnitude of ankle torque before release, which can affect recovery ability [4]. Since ankle torque scales with body height and body weight [4], we aimed to keep the ratio of ankle torque before release divided by the product of subject body height and body weight consistent across subjects and close to its average baseline value measured during quiet standing. This was achieved by using a computer monitor to provide visual feedback to the subject and investigator of foot centre-of-pressure (COP) position (an indicator of baseline ankle torque). We set a target COP position for each subject so that the ratio of COP divided by body height matched (to within $\pm$10 mm) the average value measured during quiet stance in a subset of six subjects.

During each trial, we measured the magnitude and point of application of foot–floor reaction forces and moments at a rate of 960 Hz with a force plate (model 6090H, Bertec, Worthington, OH). We also measured body segment movements with a 60-Hz, seven-camera motion measurement system (ProReflex, Qualisys Inc., Glastonbury, CT) that recorded the position of 16 markers attached to the skin and clothing overlying the right and left fifth metatarsal, lateral malleolus, lateral femoral condyle, anterior superior iliac spine, pisiform bone, radial head, acromion process, and the head and sacrum. Muscle activities in the tibialis anterior and soleus of the dominant leg were measured through surface electromyography (16-channel Myosystem 1200, Noraxon Inc.), sampled at 960 Hz. In dynamic trials, the instant of tether release was detected as the onset of a sharp decline in the tension ($\geq$2 N) measured by a load cell (Sensotec, model 31) located in series with the tether.

2.4. Data analysis

We calculated the body lean angle $\theta(t)$, defined as the angle from the vertical to a line connecting the midpoint of the two lateral malleolus markers to the midpoint of the two acromion markers [4] (Fig. 1A). We also calculated time-varying ankle plantar–flexor torque $T_a(t)$ based on the location and magnitude of vertical and horizontal components of foot reaction force:

$$T_a(t) = F_z(t)x(t) - F_x(t)z(t) - W_f x_f(t)$$

where F_x is the resultant vertical force acting on the foot (defined positive if upward), x the horizontal distance (anterior/posterior) from the ankle joint marker to where F_z acts (defined positive if anterior to the ankle), F_x the resultant horizontal force in the sagittal plane acting on the foot (defined positive if directed posteriorly), z the vertical height of the ankle joint marker above the ground, W_f the weight of the foot (0.0145 $\times$ body weight) [20], and x_f is the horizontal distance from the ankle marker to the centre-of-mass of the foot, assumed to be 0.5 of foot length [20]. In our calculation of ankle torque, we neglected inertial forces associated with angular acceleration of the feet, which have been shown to be negligible [4].

EMG recordings were high pass filtered to remove motion artifact (fourth-order Butterworth, cut-off frequency = 40 Hz), rectified, and low pass filtered (fourth-order Butterworth, cut-off frequency = 20 Hz) to determine the envelope of signal intensity. In dynamic trials, the onset of increased EMG activity in the soleus was determined as the time that signal intensity rose 3S.D.s above the mean value measured in the 500 ms preceding tether release [21–23].

From each of the three maximum recovery trials per condition, we calculated the following dependent variables: (a) the maximum recovery angle (θ_{max}) calculated as the average value of $\theta(t)$ over the 500 ms interval preceding tether release in dynamic trials, and as the maximum value of $\theta(t)$ in static trials and (b) the peak ankle torque (T_{max}) generated during balance recovery (Fig. 1B). For each subject, we also calculated the ratio of maximum dynamic recovery angle divided by maximum static recovery angle to reflect the percent decline in recovery ability due primarily to finite reaction times and rates of torque generation [4].

In dynamic trials we also calculated (1) the magnitude of ankle torque before release (T_0), calculated as the average value of $T_a(t)$ over the 500 ms preceding release (Fig. 1B); (2) the muscle response latency, defined as the interval between tether release and the onset of increased EMG activity in the soleus; (3) the electromechanical delay (EMD), defined as the interval between onset of increased EMG activity in the soleus and increased ankle torque, which was identified as the instant when $T_a(t)$ exceeded T_0 by 5 N m (always outside baseline variability); (4) the total reaction time (Δt), defined as the sum of the muscle response latency and EMD; (5) the rate of ankle torque generation following release (C), defined as the slope of a straight line joining torque–time values at the instant $T_a(t)$ exceeded T_0 by 5 N m to the instant $T_a(t)$ equaled 85% of the difference between T_0 and T_{max}; (6) the rate of ankle torque decline (D) following T_{max}, defined as the slope of a straight line joining torque–time values at the instant of T_{max} to the instant $T_a(t)$ declined by 85% of the difference between T_{max} and the minimum ankle torque during recovery. For the rate of ankle torque generation, the 85% value was chosen instead of T_{max} because it always reflected a point on the initial smooth rise of the torque–time curve, while T_{max} was sometimes offset from the initial rise due to small oscillations (secondary peaks) in $T_a(t)$. For consistency, we also chose the 85% value to define the rate of ankle torque decline. We normalized values of T_0, T_{max}, C, and D by the product of body mass (kg) multiplied by body height (m). Values of θ_{max}, T_0, T_{max}, muscle response latency, EMD, Δt, C, and D used in statistical analyses were averages over three repeated trials for each subject.

2.5. Statistics

We used one-sided independent samples t-tests to determine whether average values of θ_{max} and T_{max} in the static and dynamic trials were smaller for elderly than young. We also used one-sided independent samples t-tests to determine whether reaction time (Δt), muscle response latency, EMD, rate of ankle torque generation (C), and rate of ankle torque decline (D) in the dynamic trials, as well as the ratio of dynamic divided by static maximum recovery angle, were impaired in elderly compared to young. We used Pearson product moment correlation coefficients to test for association between maximum dynamic recovery angle, maximum static recovery angle, reaction time, rate of ankle torque generation, and peak ankle torque. In all of these tests, we used $P < 0.05$ to indicate significant differences or associations. Finally, we constructed three forward stepwise regression models to determine predictors of maximum dynamic recovery angle (θ_{max}) for (1) the entire sample, (2) young subjects, and (3) elderly subjects. The variables we entered in these models (based on the correlation coefficients) were reaction time (Δt), muscle response latency, peak ankle torque in dynamic recovery trials (T_{max}), rate of ankle torque generation in dynamic trials (C), and maximum static recovery angle (θ_{max}). Variables were entered at a significance level of $P \leq 0.05$ and removed at $P \geq 0.10$.

3. Results

3.1. Ability to recover balance

Ability to recover balance was smaller in our elderly subjects than our young subjects (Fig. 3, Table 1). In particular, the average maximum static recovery angle was 19.6% smaller in elderly than in young (mean difference = 3.2°; 95% CI: 2.0° to 4.3°, $t = 5.56$, d.f. = 39.4, $P < 0.001$), and the average maximum dynamic recovery angle was 36.1% smaller in elderly than in young (mean difference = 2.6°; 95% CI: 1.7° to 3.4°, $t = 5.84$, d.f. = 48, $P < 0.001$). In turn, the ratio of dynamic divided by static maximum recovery angle was 20.5% smaller in elderly than in young (young = 0.44 ± 0.08 versus elderly = 0.35 ± 0.12, mean difference = 0.09; 95% CI: 0.03 to 0.15, $t = 3.15$, d.f. = 40.4, $P = 0.002$).

3.2. Strength and speed of response

Differences in ability to recover balance between our young and elderly subjects reflected the combined effect of differences in strength and speed of response variables. Peak ankle torque (parameter T_{max} shown in Fig. 1B) was 4.8% smaller for elderly than young in static recovery trials, although this difference failed to reach statistical significance (mean difference = 0.045 N m/kg m; 95% CI: −0.01 to 0.10 N m/kg m, $t = 1.65$, d.f. = 33.6, $P = 0.054$), and 7.7% smaller for elderly than young in dynamic recovery trials (mean difference = 0.08 N m/kg m; 95% CI: −0.01 to 0.17 N m/kg m, $t = 1.83$, d.f. = 48, $P = 0.037$) (Table 1). In dynamic trials, it took elderly subjects longer than young subjects to initiate ankle torque development. In particular, reaction time (parameter Δt in Fig. 1B) was 27.0% slower for elderly than young (mean difference = −27 ms; 95% CI: −37 to −17 ms, $t = -5.54$, d.f. = 48, $P < 0.001$). This was due to a significantly longer muscle response latency (mean difference = −25 ms; 95% CI: −36 to −14 ms, $t = -4.51$, d.f. = 48, $P < 0.001$), as electromechanical delay was not different for elderly and young subjects (mean difference = −2 ms; 95% CI: −10 to 6 ms, $t = -0.48$, d.f. = 48, $P = 0.316$). Once torque generation was initiated, rate of

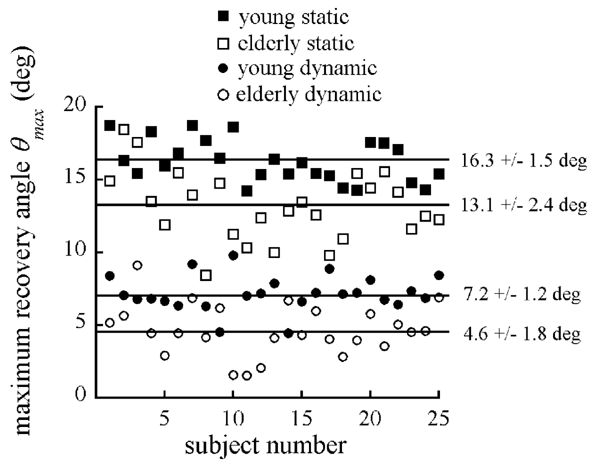

Fig. 3. Maximum recovery angles under static conditions (squares) and dynamic conditions (circles) for both young (filled symbols) and elderly (unfilled symbols) subjects. Lines show average values. The average maximum static recovery angle was 19.6% smaller in elderly than in young, and the average maximum dynamic recovery angle was 36.1% smaller in elderly than in young.

Table 1
Dependent variables by age category

Dependent variable	Young ($n = 25$)	Elderly ($n = 25$)	P-value
Maximum static recovery angle, θ_{max} (°)	16.3 ± 1.5	13.1 ± 2.4	<0.001
Maximum dynamic recovery angle, θ_{max} (°)	7.2 ± 1.2	4.6 ± 1.8	<0.001
Peak static ankle torque, T_{max} (N m)[a]	95.1 ± 17.5 (0.93 ± 0.06)	87.5 ± 20.9 (0.89 ± 0.12)	0.054
Peak dynamic ankle torque, T_{max} (N m)[a]	106.4 ± 27.3 (1.04 ± 0.15)	94.3 ± 23.4 (0.96 ± 0.16)	0.037
Rate of ankle torque generation, C (N m/s)[a]	672.6 ± 293.0 (6.60 ± 2.49)	541.2 ± 204.4 (5.57 ± 1.98)	0.057
Rate of ankle torque decline, D (N m/s)[a]	57.2 ± 27.2 (0.56 ± 0.24)	63.1 ± 23.3 (0.65 ± 0.25)	0.085
Total reaction time, Δt (ms)	100 ± 18	127 ± 16	<0.001
Muscle response latency (ms)	73 ± 22	98 ± 16	<0.001
Electromechanical delay (ms)	27 ± 14	29 ± 15	0.316

Cell entries show mean ± 1S.D.

[a] Cell entries show mean variable values followed in parentheses by mean normalized values, where normalized value = variable value/(body mass (kg) × body height (m)).

ankle torque generation (parameter C shown in Fig. 1B) was 15.6% slower for elderly than young (mean difference = 1.03 N m/s kg m; 95% CI: −0.25 to 2.30 N m/s kg m, $t = 1.62$, d.f. = 48, $P = 0.057$), and rate of ankle torque decline (parameter D shown in Fig. 1B) was 13.8% slower for young than elderly (mean difference = −0.09 N m/s kg m; 95% CI: −0.23 to 0.04 N m/s kg m, $t = -1.39$, d.f. = 48, $P = 0.085$); however, neither of these differences reached statistical significance.

3.3. Correlations between balance recovery ability, strength, and speed of response

Ability to recover balance was associated with strength and speed of response variables. The maximum dynamic recovery angle was correlated with reaction time in dynamic trials ($r = -0.657$, $P < 0.001$), with the peak ankle torque in dynamic trials ($r = 0.571$, $P < 0.001$), with the rate of ankle torque generation in dynamic trials ($r = 0.321$, $P = 0.011$), and with the maximum static recovery angle ($r = 0.669$, $P < 0.001$) (Fig. 4). These correlations persisted, with three exceptions, when young and elderly data were analyzed separately. For young subjects, maximum dynamic recovery angle was correlated with reaction time in dynamic trials ($r = -0.389$, $P = 0.027$), and with the peak ankle torque in dynamic trials ($r = 0.471$, $P = 0.009$), but not with the rate of ankle torque generation ($r = 0.235$, $P = 0.129$), or with the maximum static recovery angle ($r = 0.285$, $P = 0.084$). For elderly subjects, maximum dynamic recovery angle was correlated with reaction time in dynamic trials ($r = -0.483$, $P = 0.007$), with the peak ankle torque in dynamic trials ($r = 0.611$, $P = 0.001$), and with the maximum static recovery angle ($r = 0.515$, $P = 0.004$), but not with rate of ankle torque generation ($r = 0.254$, $P = 0.110$).

3.4. Regression analysis

The combined effect of strength and speed of response variables on ability to recover balance was shown further by regression analysis. Stepwise linear regression revealed that maximum static recovery angle, reaction time, and peak ankle torque during dynamic recovery trials together explained 62% of the variance in maximum dynamic recovery angle ($P < 0.001$). Maximum static recovery angle accounted for 45% of the variance. After accounting for maximum static recovery angle, reaction time explained another 12% of the variance, and then peak ankle torque explained a further 5%. Among young and elderly subject groups separately, maximum static recovery angle was the only significant predictor of maximum dynamic recovery

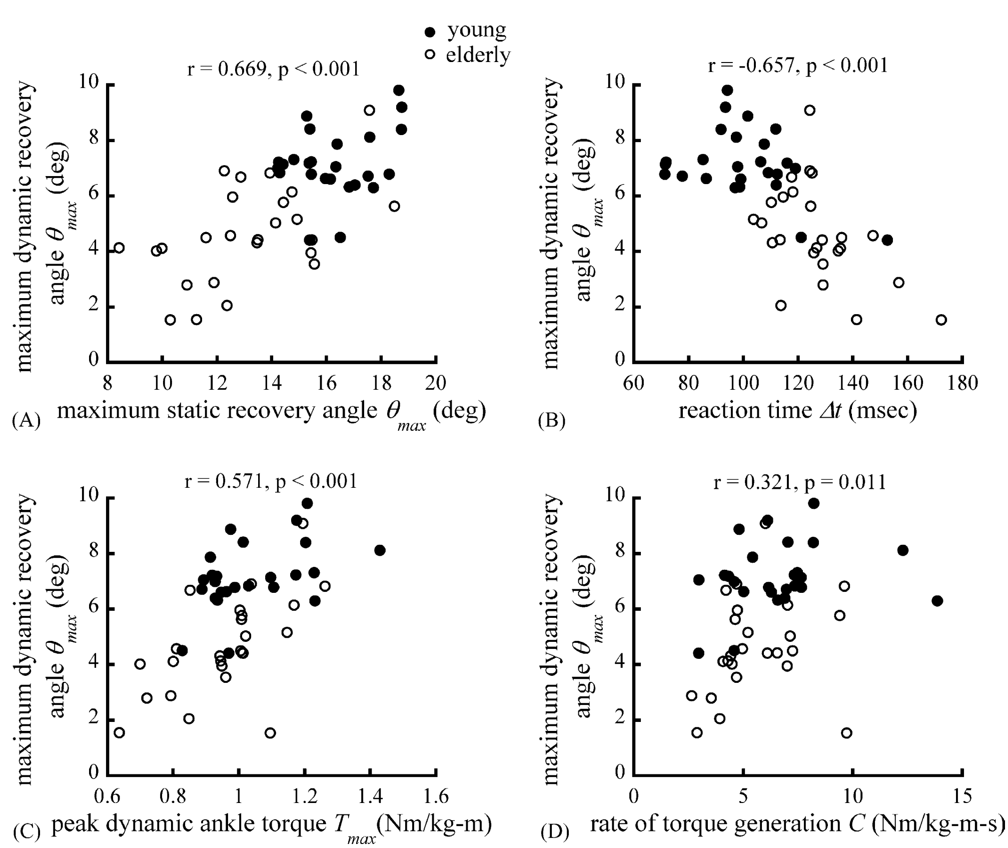

Fig. 4. Correlation between maximum dynamic recovery angle and (A) maximum static recovery angle, (B) reaction time, (C) peak ankle torque from the dynamic trials, and (D) rate of ankle torque generation, indicating that ability to recover balance is associated with both strength and speed of response variables. Young (filled circles), elderly (unfilled circles).

angle, explaining 22% ($P = 0.017$) and 37% ($P = 0.001$) of the variance for young and elderly, respectively.

3.5. Magnitude of ankle torque before release

On average, the magnitude of ankle torque before release in the dynamic trials (parameter T_0 shown in Fig. 1B), normalized for body height and body weight, was not different from normalized ankle torque during quiet stance for young (mean dynamic = 0.37 ± 0.14 N m/kg m versus mean quiet stance = 0.33 ± 0.09 N m/kg m, $P = 0.059$) or elderly (mean dynamic = 0.30 ± 0.18 N m/kg m versus mean quiet stance = 0.35 ± 0.02 N m/kg m, $P = 0.194$) subjects. Further, normalized T_0 was not different between young and elderly subjects ($P = 0.141$).

4. Discussion

The results of this study support our first hypothesis that ability to recover balance using the ankle strategy is smaller in elderly women with a history of falls than in healthy young women. This is in agreement with previous studies that reported age-related reductions in ability to recover balance with the stepping strategy [24,25]. In support of our second hypothesis, our results suggest that these differences in ability to recover balance reflect the combined effect of differences in both strength and speed of response. In particular, our results show that in comparison to healthy young women, elderly women with a history of falls have both impairments in strength (peak ankle torque and maximum static recovery angle), and impairments in speed of response (reaction time and rate of ankle torque generation). Furthermore, our correlation and regression analyses reveal that both types of decrements are associated with decreased ability to recover balance.

Our results show that the differences in reaction time between our young and elderly subjects are functionally significant, despite appearing to be small. The average difference in reaction time between young and elderly subjects in our study was 27 ms, which is similar to the 13–23 ms differences in reaction time that Wojcik et al. [24] reported for their young and elderly female subjects during stepping tasks. In the context of their stepping tasks, where recovery times averaged 500 ms, Wojcik et al. concluded that age-related delays in reaction time of 20–25 ms are not functionally significant. However, in the context of our feet-in-place balance recovery task, where recovery occurs more quickly (<200 ms) than during stepping, we have shown that these delays in reaction time are functionally significant in determining ability to recover balance. Maximum dynamic recovery angle was negatively correlated with reaction time, in agreement with predictions from mathematical models that show isolated increases in reaction time cause near linear decreases in maximum recovery angle [4]. Previous studies also show that balance recovery ability can be strongly influenced by small but diffuse neuromuscular deficits [18,26]. Moreover, reaction time strongly discriminates fallers from non-fallers [3,18].

Our results for muscle response latency and electromechanical delay suggest that neural differences in the sensing of stimuli and processing of motor commands (represented by the muscle response latency), rather than differences in muscle contraction mechanics (represented by the electromechanical delay), may govern the differences in balance recovery ability between our young and elderly subjects. Average muscle response latencies for the soleus were 25 ms slower in elderly than in young, which is similar to the 21 ms difference that Thelen et al. [27] observed between young and elderly women during rapid isometric contractions of the soleus. In contrast, we found that average electromechanical delay was not significantly slower in elderly than in young. The muscle responses that we observed had latencies longer than the monosynaptic stretch reflex but shorter than volitional reaction times, and therefore, reflect medium latency postural responses [8,28].

Localizing the neurological cause of age-related slowing of muscle response latency may help to identify the feasibility of improving reaction time in the elderly and the most promising interventions for achieving this. Previous research has established that peripheral sensory loss and afferent and efferent conduction impairment(s) result in lengthened muscle response latencies [22,23], while vestibular deficits affect the magnitude but not the timing of postural responses [29]. Alternatively, lengthened muscle response latencies in the elderly may be due to central processing deficits. These medium latency responses involve supraspinal pathways, and may be sensitive to cognitive function and the intactness of cortical and subcortical integrative systems for coordinating visual, vestibular, and somatosensory inputs. For example, studies have revealed that secondary attentional tasks impair the postural control of elderly individuals [30]. Thus, to improve our understanding of age-related declines in ability to recover balance, future research should assess how proprioceptors, cutaneous afferents, and central processing affect recovery through manipulations of somatsosensory inputs and secondary cognitive tasks.

Our results show that elderly not only take longer to generate torque in response to a postural perturbation (indicated by a longer reaction time), but once torque onset has occurred, they also develop torque at a rate that is approximately 16% slower than young. While these age differences in rate of torque generation were not statistically significant ($P = 0.057$), we consider the magnitude of the percent difference to be functionally relevant, and presumably with a larger sample size, we would have observed statistical significance. Our results for rate of torque generation are consistent with the well-documented loss of fast twitch (Type II) muscle fibers that occurs with age [5,31–33]. Furthermore, Thelen et al. [7] previously demonstrated that maximum rate of ankle torque generation was 36% slower in older females than in younger females. The discrepancy in magnitude probably relates to differences in the nature of the task: Thelen et al. measured rate of torque generation under voluntary isometric conditions while subjects lay supine, whereas we measured rate of torque generation during upright stance in response to a postural perturbation.

There are certain limitations to this study. First, we excluded individuals with major neurological, musculoskeletal, cognitive, or sensory deficits, and thus our results may have limited applicability to individuals with profound cognitive or neuromuscular impairment. Second, we conducted experiments only with women, so our results may not be applicable to men. A third limitation of the study is that the magnitude and direction of the balance perturbation were somewhat predictable to the subject. We examined only forward loss of balance and instructed subjects to recover using the ankle strategy rather than allowing them to choose between feet-in-place or stepping responses [34,35]. This is substantially different from a real-life loss of balance episode when the timing, magnitude, and direction of external perturbations are more unexpected and movement planning is more limited. However, by measuring performance in a specific balance recovery task, we were able to minimize behavioural influences and clearly identify biomechanical contributions to performance.

A fourth limitation of the study is that subject performance may have been influenced by partial reliance on the hip strategy. We visually inspected each experimental trial during data collection and repeated any trials with noticeable knee or hip flexion. In post hoc analysis, we calculated peak hip flexion rotations during recovery. For dynamic trials, these rotations ranged between 1.6° and 21.1° and averaged 6.1° in young and 9.5° in elderly. These hip flexion rotations did not associate with maximum dynamic recovery angle for either young ($r = 0.16$, $P = 0.051$) or elderly ($r = 0.01$, $P = 0.603$) subjects. Furthermore, the trend for greater hip flexion rotations in the elderly could not invalidate the main conclusions of this study. For static trials, peak hip flexion rotations ranged between 0.05° and 11.2° and averaged 1.2° in young and 1.4° in elderly. Again, these rotations also did not associate with maximum static recovery angle for either young ($r = 0.12$, $P = 0.092$) or elderly ($r = 0.01$, $P = 0.591$) subjects. This is not to say that trunk and hip muscles did not contribute to the balance recovery response. Subjects had to stabilize the hip and minimize movement between the trunk and lower extremities. Therefore, the ability to coordinate trunk and ankle muscle contractions may have limited performance in both static and dynamic conditions.

In summary, we have shown that elderly women with a history of falls have a decreased ability to recover balance with the ankle strategy when compared to healthy young women, and this difference in recovery ability is accounted for by a lengthening of muscle response latency, and decrements in rate of ankle torque generation and peak ankle torque. Thus, differences in ability to recover balance between young and elderly women associate with variables related to both strength and speed of response. These results complement growing clinical evidence that exercise programs to prevent falls in the elderly should include balance and agility training, in addition to strength training [36–38].

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