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Muscle activities used by young and old adults when stepping to regain balance during a forward fall

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Abstract

The current study was undertaken to determine if age-related differences in muscle activities might relate to older adults being significantly less able than young adults to recover balance during a forward fall. Fourteen young and twelve older healthy males were released from forward leans of various magnitudes and asked to regain standing balance by taking a single forward step. Myoelectric signals were recorded from 12 lower extremity muscles and processed to compare the muscle activation patterns of young and older adults. Young adults successfully recovered from significantly larger leans than older adults using a single step (32.2° vs. 23.5°). Muscular latency times, the time between release and activity onset, ranged from 73 to 114 ms with no significant age-related differences in the shortest muscular latency times. The overall response muscular activation patterns were similar for young and older adults. However older adults were slower to deactivate three stance leg muscles and also demonstrated delays in activating the step leg hip flexors and knee extensors prior to and during the swing phase. In the forward fall paradigm studied, age-differences in balance recovery performance do not seem due to slowness in response onset but may relate to differences in muscle activation timing during the stepping movement.

Keywords: Balance recovery; Stepping; Muscles; EMG; Falls; Aging

1. Introduction

Falls among older adults are associated with high rates of serious injury. These injuries may in part result from age-related declines in the ability to regain stable balance after a disturbance such as a trip [3]. The recovery of balance after a trip requires the sensation of a loss of balance, the planning of a recovery strategy and the execution of compensatory stepping by the musculoskeletal system. Therefore age-related changes in sensory performance, motor control, muscular strength and flexibility could all affect abilities to regain balance. Further research is necessary to ascertain how these various mechanical and neurological factors may contribute to fall injuries in older adults [15].

Significant age-related differences in the use of stepping to recover balance have been found [11,13,17,19]. For example, it was shown that the maximum forward lean angle from which balance could be regained with a single step was significantly smaller for healthy older adults than for young adults [17,19]. In those tests, maximum lean angles were highly correlated with forward step velocity. The current study expands on these results by determining if age differences in neural factors, such as muscular activation latencies and timing, could contribute to difficulties the elderly have in recovering balance by taking a rapid step upon release from a forward lean.

Age effects on muscle activities have been studied in response to platform perturbations of standing balance. It has been shown that the elderly are only slightly slower than the young in initiating active muscle

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responses following a perturbation [20,14]. In addition, older adults demonstrate significantly lower cross-correlations of activity amplitudes between synergistic muscles, and greater amounts of activity in antagonistic muscles than do young adults when regaining balance with a sway response [12,21]. The differences in antagonism may reflect a strategy of older adults relying on greater amounts of joint stiffness than young adults [21].

Age related losses in strength would likely impact abilities of older adults to step rapidly to recover balance. Older adults of ~70 years demonstrate ~20–40% losses in absolute strength (as reviewed by Doherty et al. [8]). Additionally the ability of older adults to generate joint torques rapidly is similarly diminished [16]. Thus to execute a step as rapidly as young adults, it seems that the elderly would have to adapt their muscle activation patterns, possibly using greater and longer muscle activations, to compensate for changes in muscle contraction capabilities. Based on observations of age-effects in postural perturbations and strength development, the following hypotheses were tested:

- Age does not significantly increase the latency between release from a lean and onset of muscular activities.
- Older adults will employ significantly greater muscle activities than young adults do when stepping following release from a fixed lean angle.
- The timing of muscle activities during the stepping movement will differ between young and older adults.

2. Methods

2.1. Subjects

Fourteen Young (YM: mean age 24 years, range 19 to 29 years) and twelve Old (OM: mean age 72 years, range 66 to 80 years) healthy male subjects participated. The mean (SD) heights of the YM and OM were 176.9 (7.7) and 172.6 (4.9) cm, while the mean body masses were 74.9 (11.0) and 76.1 (7.2) kg, respectively. Young subjects were recruited among University staff and students. Old were independent community-dwelling members. Potential subjects were asked to identify their dominant foot by specifying which foot they would use to kick a ball. Because the experimental setup only allowed viewing the subject from his right side, only subjects reporting that they were right foot dominant were included.

All older adults underwent a standard medical history and physical examination conducted by a geriatric clinical nurse specialist under the supervision of a geriatrician. This evaluation focused on the presence or absence of neurological or musculoskeletal abnormalities. One subject (8%) had sustained a leg fracture in the distant past. Seven subjects (58%) had occasional rare leg or back pain in the past. None of these subjects noted pain or limitation in range of motion at present. Eight subjects (67%) had decreased lower extremity reflexes and one subject (8%) had bilateral decreased vibration sense at the ankle. All twelve subjects (100%) exercised at least three times per week, involved in activities such as aerobics, biking, rowing, tennis, weight training, walking and yard work.

2.2. Forward fall experiment

A horizontal lean-control cable attached to the back of a padded pelvic belt supported the subjects while they kept their bodies essentially straight in a forward leaning posture (Fig. 1). The magnitude of the forward lean was controlled by adjusting the lean control cable length until a load cell attached to the cable indicated that it supported a specified percentage of the subject's body weight.

After a random time delay the lean-control cable was released to induce a forward fall. Subjects were instructed to attempt to regain standing balance by taking a single step forward with their right foot. All subjects wore a safety harness suspended from an overhead track. The length of the harness suspension cable was adjusted similarly for young and old, so that the subject's hands could not contact the floor in the event of a failure to restore balance upon lean-control cable release. The harness suspension cable also incorporated a load cell.

Each subject initially performed from three to six practice trials at a lean control cable load of 20 per cent of body weight. Following the practice trials, each of the subjects attempted to recover balance following releases at lean-control cable loads of 15 (three trials), 20 (one trial) and 25 (three trials) percent of body weight. After these seven trials, the supported weight was successively incremented by five percent of body weight to determine the maximum lean angle from which the subject could successfully regain balance with a single step. Trials were terminated if the subject failed twice at a given percentage of supported body weight or if he refused further trials.

Failure to recover balance was defined to occur if subjects: 1) placed 20 or more percent of their body weight on their safety harness support, or 2) took multiple steps. The harness load criterion was chosen to distinguish between the small loads exerted on the harness track by forward movement of the subject from the large loads that accompanied an assisted single step recovery. Post-hoc visual inspection of the data records showed that the 20 percent body weight criterion did discriminate between these two situations.

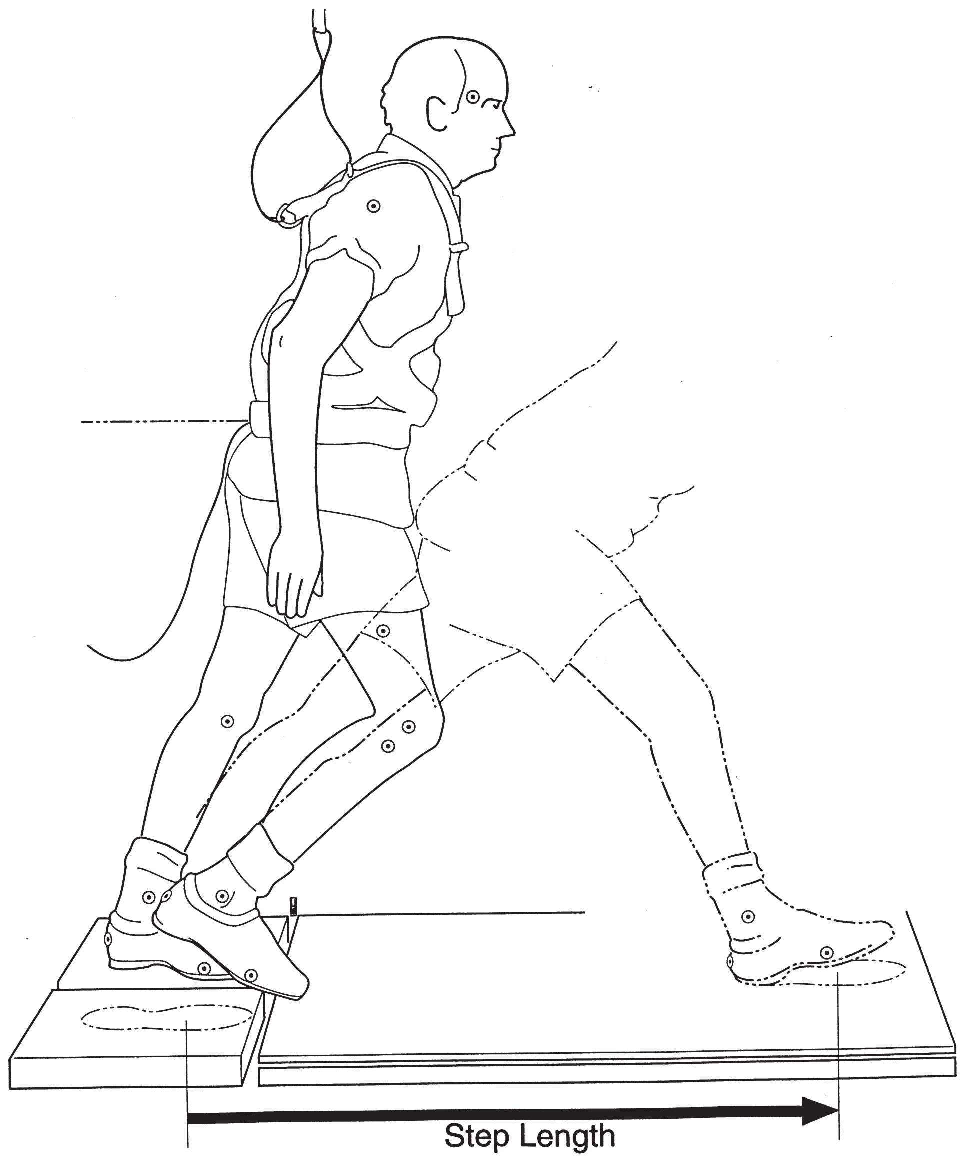

Fig. 1. Schematic diagram of experimental setup.

2.3. Experimental measurements

Data on muscular activities, body segment motions, foot/floor reactions, and lean-control and harness-support cable loads were collected synchronously. Data collection was initiated 500 ms prior to subject release, and collected for three seconds during each trial.

Myoelectric signals of 12 lower extremity muscles were measured using bipolar surface electrodes with pre-amplifiers (B and L Engineering; Santa Fe Springs, CA). The electrodes consisted of two 10 mm diameter stainless steel pads displaced 25 mm center-to-center. The electrode pre-amplifier had a gain of 330, a bandwidth of 12 to 1000 Hz, input impedance greater than 100 MΩ, and a common mode rejection ratio of 95 dB at 60 Hz. Electrodes were placed over the right and left side of the soleus, medial gastrocnemius, tibialis anterior, vastus lateralis, rectus femoris and the long head of the biceps femoris. All electrodes were placed over the mid-bellies of the muscles, oriented parallel to the underlying fiber direction. Muscle activities were collected at 2 KHz using a 64 channel, 12 Bit A/D board. The myoelectric signals were root-mean-square (rms) integrated with a moving window of 10 ms duration.

Load cells monitored lean-control and harness-suspension cable loads. Foot/floor reaction forces and moments on each side were measured using two six-component forceplates (AMTI; Watertown, MA). Step landing time was detected using a switchplate placed on the floor forward of the subject. Force data were collected at 2 KHz.

An Optotrak® (Northern Digital; Waterloo, Canada) optoelectronic motion analysis system measured body segment kinematics. Infrared emitting diodes were placed on the right leg over the lateral metatarsal, heel, lateral maleolus, lateral fibula head and lateral epicondyle; and on the thigh, midway between the knee and greater trochanter. Four diodes were placed on the medial side of the left leg; on the medial metatarsal, heel, medial maleolus and medial tibial head. Two diodes placed on the right shoulder and temple tracked trunk and head motion. Three diodes fixed to the forceplates were used to specify the location of those devices in the data set. Three-dimensional kinematic data were collected at 100 Hz. Kinematic data was used to quantify the sagittal plane movement of the body segments.

2.4. Data analysis

Stepping characteristics were evaluated by determining step liftoff time, step landing time and step length. Liftoff time was determined by finding when the vertical force under the stepping leg decreased to zero. Step landing was obtained from the switch plate. Step time was the time that the step foot was not in contact with the ground. The marker diode placed over the right metatarsal was used to calculate the length of the step taken. The average step velocity was defined as the step length divided by the step time.

Muscle latency times were defined as the interval between the release of the subject and the onset of muscular activity. Activity onsets were determined using a statistical threshold technique [5]. In this method, the activity within a 50 msec window prior to the potential onset time was used as a baseline reference. A muscle was defined to be “activated” at the end of this 50 msec window if all points in the following 25 msec window exceeded three standard deviations of the baseline activity. The 50 msec window was incremented forward in time until an onset of muscle activation was found. All myoelectric signals were also visually inspected to ensure against false positive identifications of onset times, as recommended by Di Fabio [5].

Muscle activity levels were quantified by calculating the maximum root-mean-squared (rms) myoelectric signal during any 100 ms interval between subject release and step landing. The window of maximal activity was determined by sliding the 100 ms window, in 5 ms increments, from the time of subject release up until step landing. The 100 ms window length is less than the total stepping time and is comparable to the duration of distinct bursts of muscle activity observed in the tibialis anterior and gastrocnemius muscles during forward falls similar to those of this study [7]. It is noted that similar age-differences in the magnitudes of muscle activities were found whether the signals were quantified by the rms activity during the entire stepping movement or within a 100 ms window of the response.

The timing of muscle activities of young and old adults during the stepping movement were also evaluated. For each trial, myoelectric signals were normalized to the rms activity level during the stepping response. This normalization technique allows for the evaluation of timing of muscle activities without regard to magnitudes. Young–old differences in activation timing were ascertained by using two-sided *t*-tests to compare the rms normalized activities within corresponding 10 ms intervals of the response. This was done for all 10 ms intervals between release and 500 ms after release, which is comparable to the time required to complete the first step. Due to the large number of statistical comparisons, only muscles demonstrating differences at a significance level of P=0.005 for at least three successive intervals (30 ms) were considered significant. This analysis was carried out for 25 per cent of body weight lean angles, which was the largest lean performed by all old adults.

2.5. Statistical analysis

A two-way analysis of variance (ANOVA) was used to examine the significance of differences with age and lean magnitude in step timing, step kinematics, muscular latencies and activity levels. Only data from the 15 and 25 per cent of body weight lean angles were included in the statistical analyses of the results because all subjects attempted these tests. Two-sided *t*-tests were used to determine whether there was a significant age-group differences in step timing, latencies or muscle activity levels at maximum lean angles. P values less than 0.05 were considered statistically significant.

3. Results

3.1. Maximum leans

Young adults were able to recover balance with a single step from significantly larger leans than old adults ($P<0.0001$). Maximum lean angles were 32.2° (standard deviation=3.0°) for the young adults, compared to 23.5° (2.6°) for the old adults.

3.2. Step timing and step lengths

Step liftoff and landing times each significantly decreased with increasing lean angle (Table 1). At small lean magnitudes (lean loads less than or equal to 25% of body weight), step timing and lengths were not significantly different between the young and old adults. However at maximum lean magnitudes, liftoff and landing times were each approximately 40 ms longer in the old adults. Maximum step lengths were 91 cm for the old adults, 22% shorter than those taken by the young adults. Consequently the old adults achieved a maximum step velocity that was 26% slower, on average, than the young adults.

Table 1
Mean (s.d.) timing, length and velocity of first step taken by young (Y) and old (O) adults following release from forward lean

Lean (%BW)	Liftoff time (ms)[a]		Landing time (ms)[a]		Step time (ms)		Step length (cm)[a]		Avg. step vel. (m/s)[a]	
	Y	O	Y	O	Y	O	Y	O	Y	O
15	322 (70)	315 (66)	514 (80)	546 (88)	192 (43)	230 (80)	64 (17)	72 (11)	3.45 (1.35)	3.26 (0.55)
25	255 (17)	273 (26)	444 (45)	467 (29)	189 (32)	194 (19)	82 (15)	84 (7)	4.38 (0.58)	4.32 (0.31)
Max	236 (19)	273 (27)[b]	430 (31)	471 (35)[b]	194 (20)	198 (18)	116 (12)	91 (6)[c]	6.20 (0.65)	4.61 (0.32)[c]

[a] $P<0.0005$ for lean effects
[b] $P<0.005$ for age effects at maximum lean
[c] $P<0.00005$ for age effects at maximum lean

3.3. Myoelectric latencies

Muscle latencies were independent of the lean angle for all of the muscles, and were not significantly different between the young and old adults for 10 of the 12 muscles monitored. Significantly, but only slightly, longer delays were observed in the old adults for the tibialis anterior and rectus femoris muscles of the step leg, which activated from 4 to 20 ms later in the old adults. The mean minimum latencies were independent of age, averaging less than 70 ms for both the young and old adults (Table 2). There were no significant correlations between muscle latencies and maximum lean angles of individual subjects.

3.4. Myoelectric activity levels

Maximum absolute muscular activities ranged from 241 to 848 μV, and were not significantly different between the young and old adults for any of the muscles (Table 3). At a fixed lean angle, old adults used significantly greater muscle activity than young adults did for 5 of the 12 muscles monitored. Mean absolute muscle activities were larger for old adults in the thigh muscles (vastus lateralis, rectus femoris and biceps femoris) of the stance leg, and the tibialis anterior and soleus muscles of the stepping leg.

Table 2
Mean (s.d.) muscular latency times (ms) for young and old adults. Latencies were independent of lean magnitude, so values given are the means for lean magnitudes up to 25% of body weight

Muscle[a]	Stance leg		Step leg	
	Y	O	Y	O
TA[b]	86 (13)	87 (16)	82 (10)	86 (12)
SOL	81 (11)	83 (15)	80 (9)	84 (17)
GAS	76 (16)	75 (13)	73 (13)	79 (14)
VAS	89 (16)	94 (20)	88 (16)	89 (17)
RF[b]	89 (18)	105 (27)	94 (23)	114 (38)
BF	93 (21)	99 (24)	88 (12)	84 (18)
Minimum	69 (12)	64 (11)	67 (10)	66 (9)

[a] Muscle Abbreviations: TA-tibialis anterior, SOL-soleus, GAS-gastrocnemius, VAS-vastus lateralis, RF-rectus femoris, BF-biceps femoris
[b] $P<0.05$ for age differences in step leg

3.5. Muscular activity timing

When recovering from a 25% BW lean, there were significant differences in the muscle activity timing of three stance leg muscles and two step leg muscles. (Fig. 2). At approximately the time of step foot liftoff, the old adults maintained activity levels in the stance leg soleus, gastrocnemius and biceps femoris muscles for 30 to 60 ms longer than the young adults. In the step leg, vastus lateralis activation, used to extend the knee, was delayed by 40 ms in the old adults. Rectus femoris activity, used to flex the hip and extend the knee, remained active for 90 ms longer in the old adults compared to young adults during the step swing phase.

3.6. Clinical measures

There were no significant differences in the maximum lean angle, step length, step timing or shortest muscular latencies between old subjects who were clinically diagnosed to have normal (n=4) and decreased (n=8) bilateral, lower-extremity reflexes.

4. Discussion

The current study was undertaken to determine if age-related differences in the onset, magnitude and timing of muscle activities relate to decreased abilities of old adults to recover balance by taking a single rapid step during a forward fall. The results of this study demonstrate that latency times of young and old adults are quite similar, and that the latencies are unrelated to the maximum lean angle from which a subject can recover balance. However old adults were slower in deactivating some stance leg muscles and in activating step leg muscles during the stepping movement, suggesting that age-differences in balance recovery arise during the movement.

Cross talk of myoelectric signals between neighboring

Table 3
Mean (s.d.) muscular activity levels expressed in microvolts

	Muscle[a]	15% lean		25% lean		Max lean	
		Y	O	Y	O	Y	O
Stance	TA	327 (229)	400 (136)	321 (181)	335 (114)	493 (162)	495 (141)
	SOL	270 (155)	341 (102)	371 (151)	403 (87)	655 (356)	495 (86)
	GAS†	368 (209)	402 (121)	530 (244)	529 (132)	848 (353)	650 (168)
	VAS**	119 (94)	275 (155)	177 (192)	358 (180)	578 (274)	484 (275)
	RF***,†	84 (40)	136 (69)	96 (61)	176 (69)	270 (165)	241 (119)
	BF**,†	165 (132)	329 (155)	271 (157)	397 (136)	616 (251)	521 (197)
Step	TA***,†	303 (111)	420 (151)	391 (162)	558 (148)	642 (196)	663 (184)
	SOL*	257 (163)	313 (94)	308 (128)	347 (73)	502 (200)	392 (77)
	GAS	478 (247)	508 (201)	447 (196)	487 (149)	600 (243)	620 (208)
	VAS†	210 (148)	327 (134)	343 (206)	428 (172)	588 (376)	541 (215)
	RF†	200 (100)	176 (72)	280 (31)	228 (63)	397 (167)	322 (71)
	BF	223 (155)	279 (125)	238 (94)	267 (96)	366 (180)	367 (101)

[a] ANOVA results (15% and 25% leans): Age Effects: *$P<0.05$, **$P<0.005$, ***$P<0.0005$. Lean Effects: †$P<0.05$. *T*-Test results (max leans): no significant differences between activity levels of young and old adults.

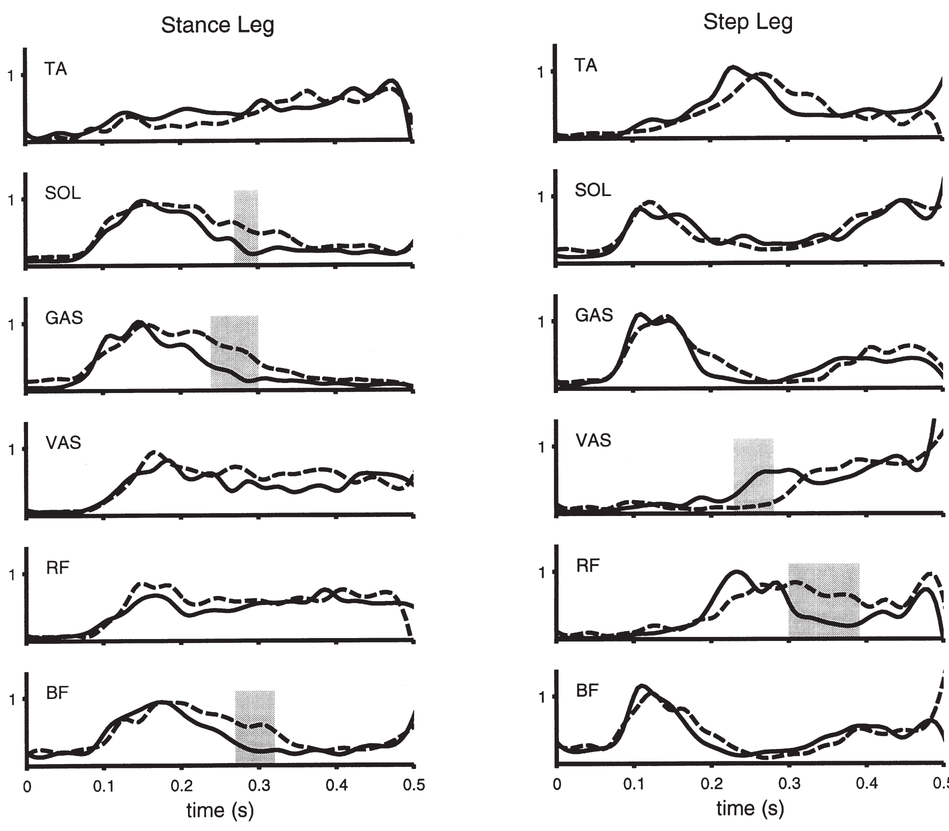

Fig. 2. Lower extremity muscle activities when stepping to recover balance following release from a 25% body weight lean. Shown are ensemble-averaged myoelectric signals for young (solid lines) and old (broken lines) adults. Regions of significant age-differences in the normalized activities are denoted by the shaded areas. See Table 2 for muscle abbreviations.

muscles may have occurred in the current study. Data from the literature indicate that cross talk magnitudes of approximately 5 to 15% can occur when bipolar surface electrodes are used, as in the current study [4], [Koh and Grabinar 1992] [10]. While cross talk could affect the absolute magnitudes of signals recorded, it probably did not influence the age-effects found since similar recording and processing techniques were used for all subjects.

Quantifying muscular latencies requires the detection of the onset, or change, in a muscular activity level. In the current study, the technique of Di Fabio [5] was used to detect when significant changes in signal level occurred. This technique has been shown to produce fairly consistent and reliable estimates of myoelectric onsets, provided that muscle activation onsets are confirmed by visual inspection [5]. The variances of the latency times were rather low, with the coefficient of variation typically being less than 20%, suggesting that latency times were fairly consistent across individuals. Muscular latency times were relatively short, with activities first detected ~70 ms after release. These latencies are faster than the ~100 ms latencies found when balance perturbations are applied to the feet during standing posture [14]. However the latencies found here are comparable to the 60 ms latencies determined by Do et al. [7] for a similar forward fall experiment, and are within the range of 60–140 ms latencies found when perturbations are applied during gait [2,9].

Muscle latencies were independent of lean magnitude and were not substantially different between the young and old adults. The independence of muscle latencies and lean magnitudes supports the idea of an invariant preparation phase prior to the stepping movement [6]. As for age effects, the step leg tibialis anterior latency was significantly longer in the old compared to the young, but only by 4 ms in the mean. This slightly longer delay probably does not have an important biomechanical effect on performance given that the additional delay represents only 1% of the 400–500 ms total step time. These results suggests that aging does not substantially slow the onset of active recovery of balance in the forward falls studied. Therefore differences in step velocity and maximum leans between young and old adults in these tests must be attributed to circumstances that occur after initiation of the active response.

To compare the relative timing of muscle activations, myoelectric signals were first normalized to the maximum rms activity recorded for that muscle during any 100 ms interval of the stepping response. The reason for not normalizing the activities to signals recorded during maximum voluntary isometric contractions is the difficulty in getting reliable maximal activity measures [22] and the problem of normalizing dynamic activity to a static measure [1]. Yang and Winter [23] showed intersubject variability in myoelectric signals was reduced substantially when signals were first normalized to either the mean or peak activity recorded during the movement, in their case a cycle of gait. This technique results in signals of similar magnitude, such that the timing of activation patterns can be compared across trials and subjects [18]. It is noted that in the current study, similar age-differences in the timing of muscle activities were found whether the signals were normalized to the rms activity during the entire stepping movement or to the max activity within a 100 ms window of the response.

The overall muscle activation patterns observed were similar for both the young and old adults (Fig. 2). Muscle activities were first seen in the gastrocnemius, soleus and biceps femoris muscles of both the step and stance leg. The gastrocnemius and soleus muscle activities help arrest passive forward rotation of the body about the ankle and in the step leg, provide a push-off prior to foot liftoff. The biceps femoris activity brakes trunk flexion about the hip and generates a flexion torque about the knee. Prior to foot liftoff, activity is seen in the step leg tibialis anterior, rectus femoris and vastus lateralis muscles. These muscles provide forces that dorsiflex the foot, flex the hip and extend the knee during the swing phase of the stepping movement. Prior to step landing, activity in the step leg gastrocnemius, soleus and biceps femoris is seen which can serve to stiffen the joints in order to absorb the impact load.

While the overall muscle activation patterns were similar, there were age differences in the timing of some muscle activities during the stepping movement. In particular, muscle activities tended to be prolonged for the older adults in the soleus, gastrocnemius and biceps femoris muscles of the stance leg. Additionally the activation of the vastus lateralis and deactivation of the rectus femoris muscles in the step leg, which flex the hip and extend the knee, were delayed. The cause of these differences may reflect a difference in recovery strategy, e.g. possibly a greater use of co-contraction to stiffen joints. Alternatively the delays in muscular deactivation and activation during the movement could be a compensation for age-differences in muscular capabilities. Many studies have documented age-related reductions in absolute strength in healthy older adults (as reviewed by Doherty et al. [8]). Additionally the ability of healthy older adults to generate joint torques rapidly is similarly diminished [16]. Thus while recovery strategies may be similar, the slightly longer muscular activation periods and larger activities may be needed by old adults to achieve the same mechanical effect as in the younger adults. Thus differences in neural activation timing may reflect a compensation for changes in mechanical capability.

In summary, old adults were less capable of taking a single rapid step to recover balance during a forward fall. Age differences in performance do not seem due to slowness in initiating a response but may relate to differences in muscle activation timing during the stepping movement.

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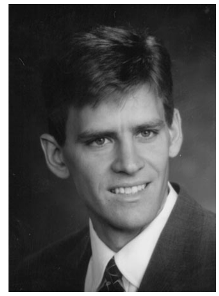

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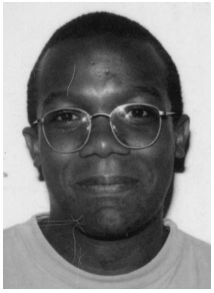

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