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Stumbling Over Obstacles in Older Adults Compared to Young Adults

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Schillings, A. M., Th. Mulder, and J. Duysens. Stumbling over obstacles in older adults compared to young adults. *J Neurophysiol* 94: 1158–1168, 2005. First published December 22, 2004; doi:10.1152/jn.00396.2004. Falls are a major problem in older adults. Many falls occur because of stumbling. The aim of the present study is to investigate stumbling reactions of older adults and to compare them with young adults. While subjects walked on a treadmill, a rigid obstacle unexpectedly obstructed the forward sway of the foot. In general, older adults used the same movement strategies as young adults ("elevating" and "lowering"). The electromyographic responses were categorized according to latencies: short-latency (about 45 ms, RP1), medium-latency (about 80 ms, RP2), and long-latency responses (about 110 ms, RP3; about 160 ms, RP4). Latencies of RP1 responses increased by about 6 ms and of RP2 by 10–19 ms in older adults compared with the young. Amplitudes of RP1 were similar for both age groups, whereas amplitudes of RP2–RP4 could differ. In the early-swing elevating strategy (perturbed foot directly lifted over the obstacle) older adults showed smaller responses in ipsilateral upper-leg muscles (biceps femoris and rectus femoris). This was related to shorter swing durations, more shortened step distances, and more failures in clearing the obstacle. In parallel, RP4 activity in the contralateral biceps femoris was enhanced, possibly pointing to a higher demand for trunk stabilization. In the late-swing lowering strategy (foot placed on the treadmill before clearing the obstacle) older adults showed lower RP2–RP3 responses in most muscles measured. However, kinematic responses were similar to those of the young. It is concluded that the changes in muscular responses in older adults induce a greater risk of falling after tripping, especially in early swing.

INTRODUCTION

Falling is a relevant health problem for many elderly. Thirty percent of community-dwelling people over 65 yr of age fall at least once a year (Tinetti et al. 1988). These falls may have a profound impact on the lives of elderly. They may lead to a decrease of activity resulting in social isolation, to serious injuries, such as fractures of the wrist and hip, and even to death (Nevitt et al. 1991; Sattin et al. 1990; Stel et al. 2004; Tinetti et al. 1995).

In risk factor studies, it has been described that an important cause of falling in community-dwelling elderly is stumbling (Blake et al. 1988). For instance, 12 to 38% of falls leading to hip fractures in older adults were caused by stumbling (Cumming and Klineberg 1994; Nyberg et al. 1996; Parker et al. 1996). Thus it is important to gain more insight into the mechanisms, which play a role in the balance problems of elderly people after tripping. Next, this insight might enable us to find ways to reduce the risk of falling.

In general, there are 2 main reactions of the body to being tripped (or stumbling strategies) dependent on the phase of the step cycle, as has been observed in prior studies with young adults (Eng et al. 1994; Schillings et al. 1999, 2000). After perturbations in the early-swing phase the "elevating strategy" was performed; after the collision with the obstacle, the foot was directly lifted over the obstacle during the perturbed swing (see Fig. 1*B*). After perturbations in the late-swing phase, subjects usually showed the "lowering strategy." After the perturbation, the ipsilateral foot was quickly placed on the treadmill without clearing the obstacle. This foot was lifted over the obstacle in the swing phase that succeeded the perturbed swing (see Fig. 1*B*).

Only a few studies have actually studied reactions of elderly after unexpected trips during walking (for review see Grabiner et al. 2002). In these studies, stumbling reactions were induced by a mechanical obstacle on a walkway, which perturbed the toe of the shoe of the swing foot during mid-to-late swing in older adults. Biomechanical variables of stumbling reactions resulting in falls (i.e., 50% of the subject's body weight supported by the safety harness ropes) were compared with the same variables during successful recoveries. Some factors at the time of perturbation increased the risk of falling, such as a high walking velocity and a more anterior center of mass of the upper body. In addition, a significantly slower placement of the tripped foot during the lowering strategy appeared to be an important factor that increased the risk of falling in older adults (Van den Bogert et al. 2002).

In these previous studies, the focus was on the factors contributing to a fall after tripping. The question remains, however, whether stumbling reactions differ between young and older adults, even if the stumbling reactions do not lead to actual falls. Are there indications that elderly react differently after being tripped?

In the leg muscles of young adults, reflex responses of various latencies have been observed after being tripped. Short-latency reflexes (latency ≅ 40 ms) were most obvious after late-swing perturbations in the perturbed leg (Schillings et al. 1999). These responses occurred simultaneously in flexor and extensor muscles and the responses were not followed by clearly observable joint angle changes. Thus it has been suggested that the short-latency responses may generate a temporary stiffening of the joints (Schillings et al. 1999). Medium-latency responses (latency ≅ 75 ms) and long-latency responses (>100 ms) occurred in muscles of both legs and were an essential part of the stumbling reactions. In midswing, the short- and medium-latency responses were similar during the 2

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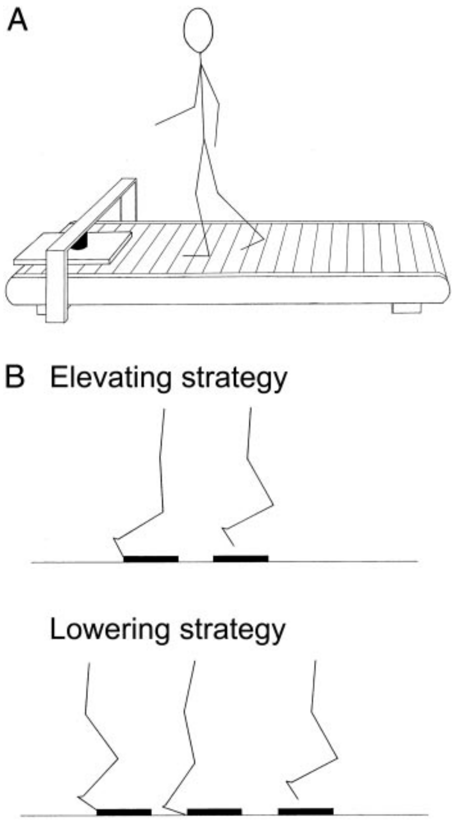

FIG. 1. *A*: experimental setup. Electromagnet (colored black) holds the obstacle above the treadmill in front of the subject's left (ipsilateral) foot. (Reprinted from *J Neurosci Methods*, Schillings AM, Van Wezel BMH, and Duysens J, Mechanically induced stumbling during human treadmill walking, 67: 11–17, 1996, with permission from Elsevier Science). *B*: schematic picture of the characteristics of the foot trajectory during the elevating and lowering strategy. Elevating strategy: after the collision with the obstacle, the foot was directly lifted over the obstacle during the perturbed swing. Lowering strategy: after the perturbation, the ipsilateral foot was quickly placed on the treadmill without clearing the obstacle; the foot was lifted over the obstacle in the swing phase that succeeded the perturbed swing (Reprinted from *J Neurophysiol*, Schillings AM, Van Wezel BMH, Mulder Th, and Duysens J, Muscular responses and movement strategies during stumbling over obstacles, 83: 2093–2102, 2000, with permission from The American Physiological Society.)

stumbling strategies (elevating and lowering). Possibly this initial aspecific response provides the CNS some time to integrate information from sensory receptors and supraspinal sources to make an appropriate decision about the final behavioral strategy, which was mainly determined by the responses occurring later than about 100 ms after the perturbation (Schillings et al. 2000). The amplitudes of the short-, medium-, and long-latency responses were dependent on the phase of perturbation in the step cycle (Schillings et al. 2000).

For stumbling it is unknown whether elderly have longer response latencies and lower response amplitudes than those of young adults. Some evidence pointing in that direction comes from studies with another type of unexpected perturbations of older adults during walking, that is, induced slips. Tang et al. (1998, 1999) described longer response latencies and lower response amplitudes after slipping at heel strike in older adults compared with young adults. On the basis of these studies, it is hypothesized that during stumbling older adults may also have longer electromyographic response latencies and lower response amplitudes than those of young adults. The aim of the present study was to induce trips during walking in older adults, and to compare the stumbling strategies and the concurrent electromyographic and kinematic responses of older adults with the responses of young adults.

METHODS

Eight older healthy, community-dwelling adult subjects (5 male, 3 female) aged between 60 and 73 yr (mean age 65 yr) participated in the experiment. Subjects reported to be free from neurological or motor disorder and had no known history of repeated falling. They were all active elderly and reported to be capable of walking continuously for at least 2 h. The experiments were carried out in conformity with the declaration of Helsinki for experiments on humans. All subjects gave informed consent and the study was approved by the local ethical committee.

Experimental setup

Because a detailed account of the experimental setup can be found in Schillings et al. (1996), only a short description is given here. While subjects walked on a treadmill (speed 4 km/h), an obstacle (length, width, and height: 40.0, 30.0, and 4.5 cm, respectively; weight 2.2 kg) was held by an electromagnet above the treadmill about 1 m in front of the subject (see Fig. 1*A*). To induce perturbations, the obstacle was dropped unexpectedly on the belt thereby obstructing the forward sway of the left (ipsilateral) leg. Release of the obstacle occurred at a predetermined delay after ipsilateral or contralateral heel strike. A pressure-sensitive strip attached to the front of the obstacle was used to measure the time at which the foot hit the obstacle. In the thin flexible shoes, the toes were covered with a piece of cotton to protect them during the collision. The subjects wore a pair of glasses, which blocked downward sight (and thus blocked the view of the obstacle). Earplugs eliminated most of the sound produced by the obstacle landing on the treadmill. In addition, the sound was masked by music through headphones. As a result of these measures, subjects were not able to perceive the obstacle before the collision with the foot. Subjects were instructed to keep the same position on the treadmill before the perturbation, but after the collision they were free to react without restrictions. The subjects wore a safety harness, fixed to a safety brake on the ceiling that would hold the subject and stop the treadmill in case a subject should fall. The safety brake was activated when a force of $>$120 N was exerted. The harness was loosely suspended so that it did not provide extra stability during the experiment.

Data sampling

Bipolar surface electromyogram (EMG) activity of the biceps femoris (BF), rectus femoris (RF), tibialis anterior (TA), and soleus (SO) of both legs was measured. Laterally placed goniometers were used to measure the joint angles of the knee and ankle of the ipsilateral leg. Thin insole foot switches measured foot contact with the treadmill. Data were sampled in a time interval starting 100 ms before triggering the electromagnet and lasting for 2,100 ms. For the control trials the same intervals were sampled but no obstacle was dropped after the trigger. The EMG was (pre-) amplified (by a factor in the order of 10^4 to maximally 10^6), high-pass filtered ($>$3 Hz), full-wave rectified, low-pass filtered ($<$300 Hz), AD-converted (500 Hz), and stored on hard disk along with the signals of the goniometers, foot switches, and pressure-sensitive strip. In addition, the subjects were recorded on video (25 Hz) during the experiment.

Experimental protocol

The experiment consisted of 3 parts. The first and second part were preparations for the main experiment (third part). In the first part, subjects walked on the treadmill for ≥5 min to get used to the treadmill (unperturbed walking). In the second part (20 min), the obstacle perturbed the walking pattern in phases spread all over the step cycle. The computer triggered the electromagnet to drop the obstacle on the treadmill after fixed delays (0, 40, 80, . . . , 600 ms) after heel strike (in total 32 delays). Each delay condition was randomly applied only once, consistent with the protocol used for the young adults. The aim of this part of the experiment was to select the delay conditions in which stumbling reactions were evoked in early (5–25%, time of obstacle contact with respect to control swing duration), mid (30–50%), and late swing (55–75%). The 3 selected delay conditions were used in the third part of the experiment.

In the third part of the experiment (30 min), stumbling reactions were repeatedly and randomly introduced during early, mid, and late swing to construct averages. On average 8 trials (minimal 5 trials) were obtained for each phase of perturbation. The responses during these perturbed cycles were compared with unperturbed control trials (n = 10 to 15) obtained in between the perturbation trials (perturbation-free period between trials >10 s).

Data analysis

To obtain the averaged responses, only trials in which the collision of the foot with the obstacle occurred during early (5–25%) and late swing (55–75%) were analyzed (the midswing data were not further analyzed because the number of trials per strategy was too small to allow a quantitative analysis). For each muscle of each subject separately, the stumble responses occurring in the same phase of the step cycle were averaged. In addition, the corresponding control trials were averaged. Subsequently the averaged control activity was subtracted from the averaged stumbling trials.

To quantify the amplitudes of the responses, the mean EMG activity was calculated in the period between the beginning and end of the response. For this purpose, windows were set around the individual response peaks occurring within the first 200 ms (see Schillings et al. 2000); windows were set around responses with short latencies (about 45 ms, RP1), medium latencies (about 80 ms, RP2), and 2 windows for responses with long latencies (about 110 ms, RP3; and about 160 ms, RP4, respectively). When a response did not have the shape of a sharp peak, an extrapolation was made on the basis of other trials or conditions of the same subject in which such peaks were clearly detected. In this way it was possible to discern peaks with latencies that corresponded well with those seen in young adults (Schillings et al. 2000). The RP1–RP3 responses (occurring before about 150 ms) are considered to be reflex responses, whereas the RP4 responses (occurring after about 150 ms) might be under voluntary control. The latter conclusion is based on the findings of a study of Hase and Stein (1998), in which subjects were instructed to stop walking as soon as they got a cue by electrical stimulation of the superficial peroneal nerve. It was found that the earliest voluntary changes in EMG activity of leg muscles occurred 150–200 ms after stimulation.

To enable a proper intersubject comparison of the response amplitudes, the resulting data of each muscle were normalized with respect to the maximal EMG activity during the control step cycles. To obtain the mean response of the whole group, the normalized responses of all older adults were averaged. This type of analysis was performed on the 4 response peaks. The Wilcoxon matched-pairs signed-ranks test ($P < 0.05$) was used to test whether the response amplitudes during stumbling were significantly different from the control EMG activity.

Video analysis

To find out whether a lengthened swing duration of the perturbed leg also implied a lengthened step distance, video recordings were analyzed. For both control trials and stumble trials, the distance between the 2 feet at the time of ipsilateral heel contact was determined using video analysis of marks on the treadmill belt (distance between 2 marks = 6.0 cm). In addition, for the stumble trials it was determined whether a secondary contact of the ipsilateral foot with the obstacle occurred.

Comparison of the data of older adults with young adults

The results of the present study were compared with the stumbling reactions of young adults, which were mostly described in a prior study (Schillings et al. 2000). The number of young subjects was 8 (5 male, 3 female, mean age 27 yr). The latencies of the short-latency responses and medium latencies were determined automatically for the average subtracted EMG responses of all older adults separately, and of all young adults separately. First, the SD of the nonperturbed subtracted EMG activity was calculated in the last 100 ms before the perturbation. Second, the latencies of the stumble responses were defined as the time at which the EMG amplitude exceeded 3 times the SD and stayed above this level for at least 4 succeeding samples (about 8 ms in duration). In some cases where the EMG activity did not return under the 3SD level after the short-latency response, the onset of the medium-latency response was defined as the onset of the subsequent increase of EMG activity. To compare the response latencies of older and young adults, the Wilcoxon rank-sum test ($P < 0.05$) was used. When the EMG activity did not meet the criteria (EMG level >3SD, duration >8 ms), the data were treated as missing data. With this automatic detection criterion, it was not possible to distinguish reliably between the medium- and long-latency responses because in the majority of trials the EMG activity did not return under the 3SD level. Thus the latencies of the RP3 and RP4 responses could not be compared between the 2 age groups.

The EMG amplitudes calculated from the 4 time windows in older adults (see *Data analysis*) could be compared with the responses in young adults because the same method was used to quantify the amplitudes of responses in young adults (Schillings et al. 2000). The Wilcoxon rank-sum test ($P < 0.05$) was used to compare the response amplitudes of the ipsilateral biceps femoris (iBF), ipsilateral rectus femoris (iRF), ipsilateral tibialis anterior (iTA), ipsilateral soleus (iSO), and contralateral biceps femoris (cBF) in the 2 groups.

RESULTS

Normal walking

All older adults were capable of walking at the speed of 4 km/h imposed by the treadmill. The average step-cycle duration during the control trials, in which the walking pattern was not perturbed, was 1,088 ms (SD 79 ms). This was similar to the step-cycle duration found in young adults (step cycle 1,120 ms, SD 65 ms). The average duration of the swing phase in older adults was 438 ms (SD 32 ms; young adults mean 454 ms, SD 28 ms) and of the stance phase was 650 ms (SD 48 ms; young adults mean 666 ms, SD 41 ms). These control step parameters were not significantly different between older and young adults, as determined with the Wilcoxon rank-sum test.

Stumbling strategies

All older adults were able to restore their walking cycle by making a stumbling movement after mechanical obstruction of the forward swinging foot. None of the subjects completely lost balance after being tripped. In principle, a loss of balance could be identified by a stop of the treadmill, because a large force (>120 N) on the safety harness (attributed to the subject's weight) would activate the safety brake. However, this never happened during the stumbling experiments. Similar to

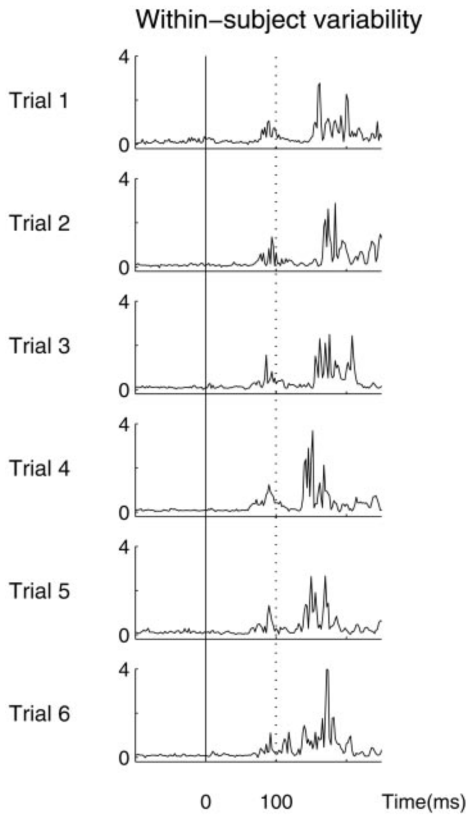

FIG. 2. Within-subject variability of the responses. *Traces*: ipsilateral biceps femoris (iBF) electromyogram (EMG) activity (normalized to the maximum background activity) during individual trials of one typical subject after early-swing perturbations.

the stumble responses of young adults, all older adults used 2 different types of stumble strategies. The choice for one of the 2 strategies depended on the timing of the perturbation in the step cycle. The stumbling responses after perturbations in the same phase of the step cycle were quite reproducible within the same subject, as can be observed in Fig. 2. All older adults showed elevating strategies after perturbations in early swing (5–25%) and lowering strategies in late swing (55–75%). This is in agreement with the strategies performed by young adults, who always showed elevating strategies in the first 35% of the swing phase and lowering strategies after 52%.

Figure 3 shows the reactions during the early-swing elevating strategy and the late-swing lowering strategy for one (typical) older subject. Soon after the early swing perturbation, the ipsilateral knee started to flex to lift the foot over the obstacle (Fig. 3*A*). Knee flexion was assisted by increased iBF activity, which was followed by increased iRF activity assisting the subsequent knee extension. Only about 20 ms after the perturbation, the ankle showed (passive) plantar flexion followed by (dorsi-) flexion, presumably related to the increased activity in iTA. Because of these reactions, the foot trajectory of the ipsilateral foot was changed, which led to a lengthened swing-phase duration (see iFoot).

After the collision in late-swing perturbations, the ipsilateral foot was first placed on the treadmill, before crossing the obstacle. To place the foot, responses in iBF and iRF occurred simultaneously (see Fig. 3*B*). In this way the forward swing could be decelerated (iBF) and the knee extended (iRF). In the lower leg, iTA responses were followed by responses in iSO, which could take up body support after foot placement.

Although some differences are apparent, it is striking to see that these patterns of muscle and movement responses of the elevating and the lowering strategy were generally quite similar to the response patterns observed in young adults (Schillings et al. 2000). The similarity of the muscle response patterns is apparent in Figs. 4 and 5, showing the average EMG responses of all older subjects along with the average responses of all young subjects.

Nevertheless, a closer look at the data revealed several differences. In Figs. 4 and 5 it can be seen that in both phases of perturbation, the average EMG responses of several muscles in older adults had longer latencies and lower amplitudes than those in young adults (for a clear example see iRF). These

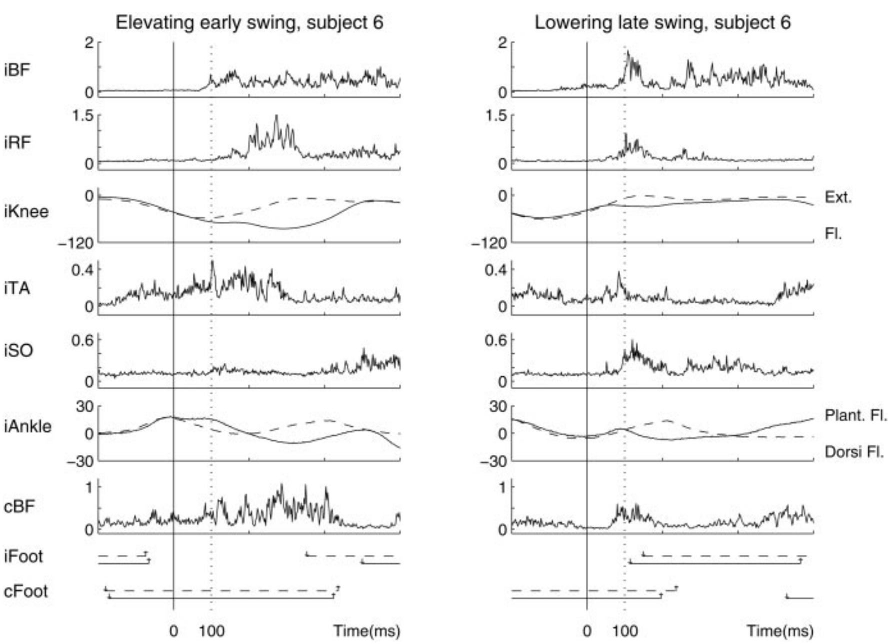

FIG. 3. Typical average subtracted EMG responses and joint angle changes for subject 6. Responses during the early swing elevating strategy (n = 5 trials) and responses during the late swing lowering strategy (n = 5 trials). Averaged subtracted EMG responses (mV) are shown for the ipsilateral biceps femoris (iBF), rectus femoris (iRF), tibialis anterior (iTA), and soleus (iSO) and for the contralateral biceps femoris (cBF). Joint angle changes (Degrees; not subtracted) are shown for the ipsilateral knee (iKnee) and ankle (iAnkle). Angle at standing position is zero. Two *bottom traces* show stance phases of the ipsilateral (iFoot) and contralateral foot (cFoot). Goniometer and foot signals: solid lines indicate stumble responses; dashed lines indicate control data (n = 15). *Time 0* is the time the foot collided with the obstacle. Ext., extension; Fl., flexion; Plant. Fl., plantar flexion; Dorsi Fl., dorsiflexion.

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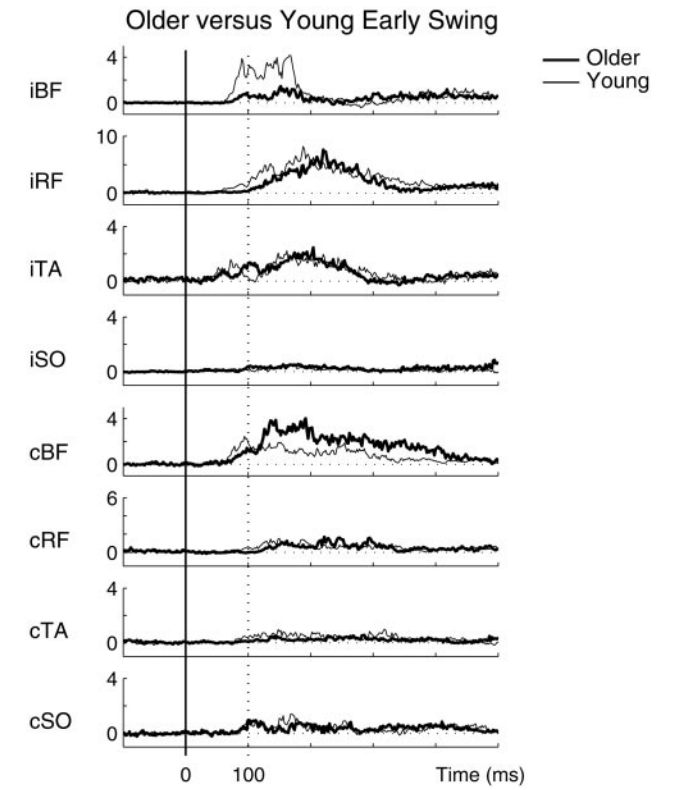

FIG. 4. EMG responses during the early swing elevating strategy for older and young adults. Averaged normalized subtracted responses of all older adults (n = 8) are shown by heavy lines and of all young adults (n = 8) by thin lines. EMG responses (normalized to the maximum background activity) are shown for the ipsilateral and contralateral biceps femoris (iBF; cBF), rectus femoris (iRF; cBF), tibialis anterior (iTA; cTA), and soleus (iSO; cSO). *Time 0* is the time the foot collided with the obstacle.

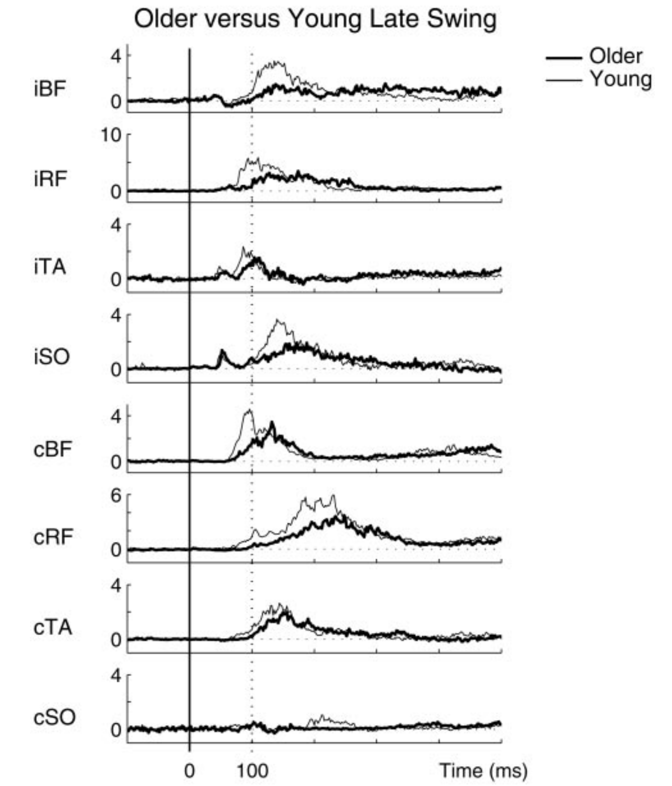

FIG. 5. EMG responses during the late swing lowering strategy for older and young adults. Averaged normalized subtracted responses of all older adults (n = 8) are shown by heavy lines and of all young adults (n = 8) by thin lines. EMG responses (normalized to the maximum background activity) are shown for the ipsilateral and contralateral biceps femoris (iBF; cBF), rectus femoris (iRF; cBF), tibialis anterior (iTA; cTA), and soleus (iSO; cSO). *Time 0* is the time the foot collided with the obstacle. Same format is used as in Fig. 4.

differences in response latency and response amplitude were further analyzed quantitatively.

Muscle response latencies

First, short-latency responses (latency ≅ 45 ms) were small in early swing (Fig. 4). Thus the averages of the responses were rarely larger than the 3SD level used to automatically detect the response latencies in each subject. In late swing, however, the short-latency RP1 responses in iRF and iSO did exceed the 3SD level in 5 subjects (in both young and older adults). The response latency in iRF was not significantly lengthened in older adults (iRF average latency: young 41 ms, SD 4.1 ms; older 44 ms, SD 4.6 ms). In iSO, however, the latency of the short-latency response was significantly increased by 6 ms in older adults (average latency young: 43 ms, SD 2.3 ms; older: 49 ms, SD 3.3 ms, Wilcoxon rank-sum test).

Second, it can be observed that the latencies of the main responses (mostly medium-latency responses with latencies of about 75 ms) of older adults were mostly longer than the latencies observed in young adults (see Figs. 4 and 5). The latencies of the medium-latency responses were determined for all subjects separately, allowing a statistical analysis (see Table 1). In early swing a significant difference was found between older and young adults in 3 muscles: iBF, iTA, and iSO. In these muscles, the response latencies of older adults compared with those of young adults increased by 10, 19, and 17 ms, respectively. In late swing, a significantly longer latency was observed as well; in the cBF muscle, the average latency in older adults was 10 ms longer than that in young adults.

Muscle response amplitudes

To quantitatively compare the amplitudes of the responses, the average EMG activities within certain windows of interest

TABLE 1. *Latencies of medium-latency responses of older versus young adults*

	Early Swing		Late Swing	
	Older	Young	Older	Young
iBF	78 (±10; 7)*	68 (±5; 8)	93 (—; 2)	80 (±12; 3)
iRF	72 (—; 2)	76 (±7; 7)	77 (±5; 3)	77 (±10; 8)
iTA	88 (±5; 4)*	69 (±3; 5)	82 (±8; 3)	78 (±10; 8)
iSO	95 (±4; 4)*	78 (±8; 5)	90 (±5; 5)	82 (±7; 6)
cBF	77 (±8; 3)	72 (±10; 8)	78 (±6; 6)*	68 (±8; 8)
cRF	—	87 (±6; 3)	78 (—; 1)	73 (±8; 7)
cTA	82 (—; 1)	80 (—; 1)	89 (—; 2)	80 (±12; 4)
cSO	—	95 (±2; 3)	—	64 (—; 1)

Mean latencies in milliseconds after perturbation of older versus young adults (±SD; n = number of subjects included in average). Note that in older adults the number of subjects with observed medium-latency responses >3SD was smaller than in young adults (and was sometimes zero, as indicated with —). *indicates a significant difference in response latency between older and young adults (Wilcoxon rank-sum test, $P < 0.05$). i, ipsilateral; BF, biceps femoris; RF, rectus femoris; TA, tibialis anterior; SO, soleus; c, contralateral.

TABLE 2. *Mean onset and end of windows around the 4 response peaks*

Muscle	Function	RP1	RP2	RP3	RP4
iSO	Ankle plantar flexion	O: 47–65 (±4)	O: 80–101 (±5)	O: 112–143 (±6)	O: 159–194 (±8)
		Y: 42–67 (±3)	Y: 76–100 (±4)	Y: 112–147 (±6)	Y: 162–197 (±12)
iTA	Ankle dorsiflexion	O: 47–65 (±6)	O: 83–100 (±3)	O: 107–139 (±9)	O: 158–195 (±6)
		Y: 41–64 (±3)	Y: 78–105 (±7)	Y: 120–144 (±3)	Y: 164–197 (±3)
iBF	Knee flexion	O: 36–50 (±4)	O: 80–100 (±5)	O: 107–143 (±3)	O: 158–194 (±6)
	Hip extension	Y: 34–53 (±4)	Y: 71–98 (±5)	Y: 103–143 (±3)	Y: 159–200 (±8)
iRF	Knee extension	O: 51–71 (±16)	O: 80–98 (±5)	O: 108–147 (±7)	O: 162–195 (±7)
	Hip flexion	Y: 39–63 (±4)	Y: 72–94 (±6)	Y: 104–140 (±5)	Y: 160–198 (±7)
cBF	Knee flexion	—	O: 77–99 (±7)	O: 113–145 (±6)	O: 161–196 (±7)
	Hip extension	—	Y: 74–104 (±5)	Y: 118–151 (±10)	Y: 164–201 (±14)

Mean onset and end of windows, which were set around the 4 response peaks (RP1–RP4) of older (O) and young (Y) adults in the muscles iSO, iTA, iBF, iRF, and cBF. These values are means of all subjects (n = 8 for both groups) in milliseconds after perturbation. Number in parentheses is ±SD of the window onset. The movement functions of the muscles in the saggital plane are indicated in the second column. i, ipsilateral; SO, soleus; TA, tibialis anterior; BF, biceps femoris; RF, rectus femoris; c, contralateral.

were compared between older and young adults (see Table 2 and Schillings et al. 2000). Figures 6 and 7 show the average EMG amplitudes of the older adults compared with those of young adults for early-swing and late-swing perturbations, respectively. The dark bars indicate the background activity during unperturbed walking. No significant differences were found between the background EMG activity in older and young adults, as determined by the Wilcoxon rank-sum test. When the amplitude of the stumble response was significantly different from the background activity an asterisk is shown below the bar (Wilcoxon matched-pairs signed-rank test, $P < 0.05$). A larger asterisk in the center above the top of the bars indicates a significantly different response amplitude between older and young adults (Wilcoxon rank-sum test, $P < 0.05$).

Significant differences were usually attributed to smaller responses in older adults (compared with those in young adults). In early swing, this can clearly be observed for iBF and iRF (see Fig. 6). The amplitudes of the medium-latency responses (RP2) of the iBF and iRF as well as the long-latency responses (RP3) of iBF were significantly lower in older adults. At most older adults increased their iBF activity to a level beneath the maximum EMG activity during normal walking (RP2: about 0.6; RP3: about 0.5; values indicate subtracted normalized amplitudes; see Fig. 6). However, young adults increased their iBF activity to approximately 3 times the maximum EMG activity during normal walking (amplitude of RP2: 2.4; RP3: 3.1). Similarly, in late swing significantly lower amplitudes were found in older adults. The medium-latency responses (RP2) of iTA and cBF were smaller for older than for young adults (iTA, RP2, 0.9 in older adults vs. 1.8 in young adults; cBF, RP2, 1.0 in older adults vs. 3.2 in young adults). The long-latency responses (RP3) of iSO and iBF were also smaller in older adults (see Fig. 7).

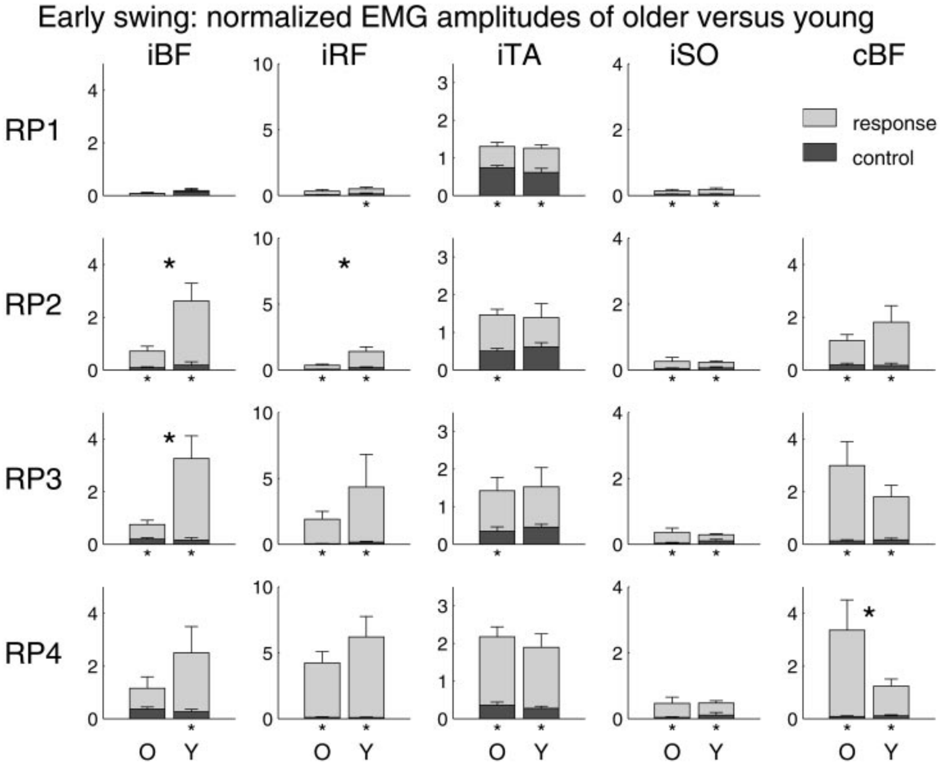

FIG. 6. Mean normalized EMG amplitudes of the 4 response peaks (RP1–RP4) in iBF, iRF, iTA, iSO, and cBF of older (*O*) and young (*Y*) adults during the early swing elevating strategy. Dark bars show the normalized background activity; light bars show the normalized EMG response amplitudes during the stumbling reactions (normalization with respect to the maximum background locomotor activity). In this way, the part of the light bars above the dark bars indicates the amplitude of the subtracted responses. SE of the control activity and the subtracted response activity is shown by error bars. Averaged responses were calculated by averaging the mean responses of all subjects: n = 8 for both older and young adults (number of trials per subject: 5–10, except for one subject with 3 trials). * below the bars, pooled average response amplitudes of all subjects were significantly different from the averaged control activity (Wilcoxon matched-pairs signed-ranks test, $P < 0.05$). * above the bars, pooled average response amplitudes of all older adults were significantly different from the pooled average response amplitudes of all young adults (Wilcoxon rank-sum test, $P < 0.05$).

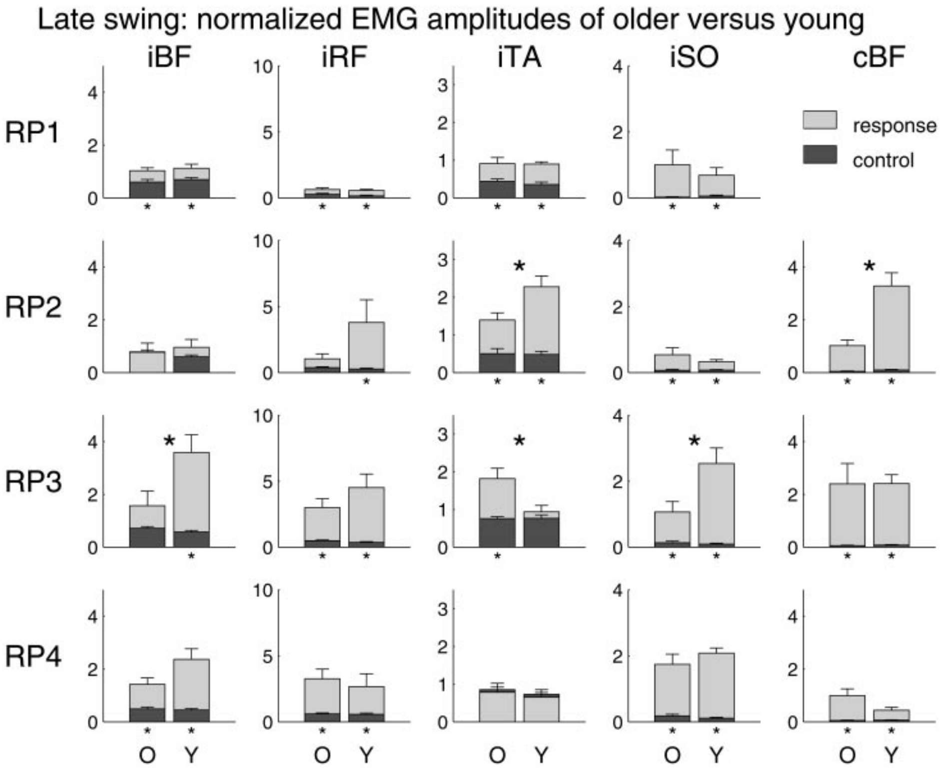

FIG. 7. Mean normalized EMG amplitudes of the 4 response peaks (RP1–RP4) in iBF, iRF, iTA, iSO, and cBF of older (*O*) and young (*Y*) adults during the late swing lowering strategy. Dark bars show the normalized background activity; light bars show the normalized EMG response amplitudes during the stumbling reactions (normalization with respect to the maximum background locomotor activity). In this way, the part of the light bars above the dark bars indicates the amplitude of the subtracted responses. In case the response was suppressive, the dark bar was larger than the light bar (see for example, iTA, RP4). SE of the control activity and the subtracted response activity is shown by error bars. Averaged responses were calculated by averaging the mean responses of all subjects: $n = 8$ for both older and young adults (number of trials per subject: 5–10). Same format is used as in Fig. 6.

Whereas RP2 responses were always smaller (or equal) in older adults compared with those in young adults, for the later responses (RP3 and RP4) the inverse was sometimes true. An example is the cBF activity of the long-latency responses (RP4) in early swing. The normalized amplitude of this response was 3.3 in older adults, which was significantly larger than the RP4 response of 1.1 in young adults. Also in late swing the RP3 response of iTA in older adults (1.1) was significantly larger than that in young adults (0.2).

Kinematics of the early-swing elevating strategy

The increased response latencies as well as the lower response amplitudes in older adults compared with those in young adults might influence the kinematics of the stumble responses. After early-swing perturbations, for example, it is clear that the iBF response (latency older adults 78 ms; young adults 68 ms) was smaller in older than in young adults (see Figs. 4 and 6). The iBF response contributed to the knee flexion of the ipsilateral leg to step over the obstacle. To see whether this diminished activity resulted in changes of the kinematics during the early-swing elevating strategy, the average joint angle changes were investigated (see Fig. 8).

Both older and young adults increased the knee flexion during the perturbed swing (solid lines) compared with the knee flexion during normal walking (dotted lines) to lift their foot directly over the obstacle. Older adults showed less increase of the knee flexion than that in young adults (Wilcoxon rank-sum test, $P = 0.32$). Although the first part of the iTA activity (between about 70 and 120 ms) showed some difference (see Fig. 4), the ankle joint changes of older and young adults were similar during the early-swing elevating strategy.

What do these differences in EMG activity and joint movements mean for the duration of step parameters in early swing? During the early-swing elevating strategy, both young and older adults lengthened the duration of the ipsilateral swing phase to step over the obstacle (see Fig. 8, solid lines in *bottom trace*). However, older adults lengthened their swing phase by

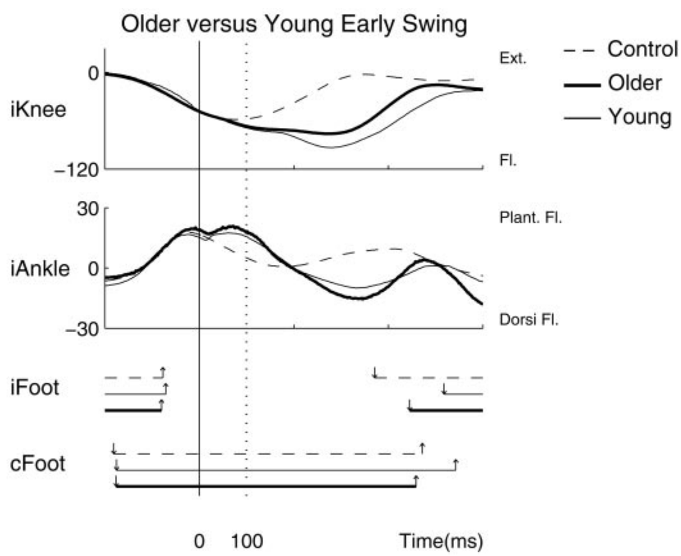

FIG. 8. Pooled joint angle changes (Degrees; not subtracted) during the early swing elevating strategy for the ipsilateral knee (iKnee) and ankle (iAnkle) of all older and all young adults. Angle at standing position is zero. Two *bottom traces* show stance phases of the ipsilateral (iFoot) and contralateral foot (cFoot). Thin lines indicate pooled stumble responses of all young adults ($n = 8$); heavy lines indicate pooled stumble responses of older adults ($n = 4–8$); dashed lines indicate control data (average of pooled controls of older and young adults). *Time 0* is the time the foot collided with the obstacle. Ext., extension; Fl., flexion; Plant. Fl., plantar flexion; Dorsi Fl., dorsiflexion.

on average 69 ms (which was about 16% of normal swing-phase duration, SD 40 ms, Wilcoxon matched-pairs signed-ranks test, $P < 0.01$), whereas young adults lengthened their swing phase by 128 ms (which was about 29% of normal swing phase duration, SD 30 ms; see Schillings et al. 2000). The lengthening of the swing phase was significantly reduced in older adults (Wilcoxon rank-sum test, $P < 0.01$).

The same can be observed for the contralateral stance phase. Although older adults did not significantly change the contralateral stance phase (on average the contralateral stance phase was shortened by 11 ms, about 2%, SD 46 ms), young adults significantly lengthened the contralateral stance phase by 81 ms (about 12%, SD 41 ms; Schillings et al. 2000). The difference in the duration of the contralateral stance phase between older and young adults was significant (Wilcoxon rank-sum test, $P < 0.01$).

Failed obstacle clearance during the elevating strategy

On the video recordings, the number of trials in which the subjects had secondary contact with the obstacle was counted. Secondary contact occurred because the subject landed with the foot on top of the obstacle, after an attempt to step over it. In older adults, in 10 out of 56 trials a secondary contact was observed. For 5 of the older adults the secondary contact occurred in one single trial, for one subject it happened in 2 trials, for another in 3 trials. For only one older subject no secondary contact occurred. In the young subjects, merely one secondary contact was observed in the whole group (total number of trials 56).

The risk to step on the obstacle increases when a shorter step distance is made. Therefore on the video recordings it was determined whether the step distance during the elevating strategy was shorter than the step distance during normal walking. The video recordings revealed that during the elevating strategy, older adults shortened their step *distance* in 73% of all trials (although they significantly lengthened their sway *duration*). In contrast, young adults shortened their step distance in only 30% of all trials. Indeed, it was found that most secondary contacts of older adults occurred in the trials in which the step distance was shortened; 9 out of 10 secondary contacts of older adults were made in trials with a shortened step distance.

Kinematics of the late-swing lowering strategy

Comparing the late-swing lowering responses of older adults and young adults, the EMG activity of all muscles reached lower maximum amplitudes in older adults (Fig. 5). The joint movements in the ipsilateral leg of older adults, however, showed relatively small changes in older adults with respect to young adults (see Fig. 9). Possibly older adults used other strategies to make the same joint movements. For example, it might be possible that they recruited other muscles, which were not recorded.

The average latency of foot placement of the ipsilateral foot after the perturbation in older adults was 151 ms (SD 60 ms), which was slightly longer than the latency of foot placement in young adults (125 ms, SD 35 ms; Wilcoxon rank-sum test, $P = 0.69$). Because foot placement of the ipsilateral foot is considered to be actively controlled by iRF for knee extension and

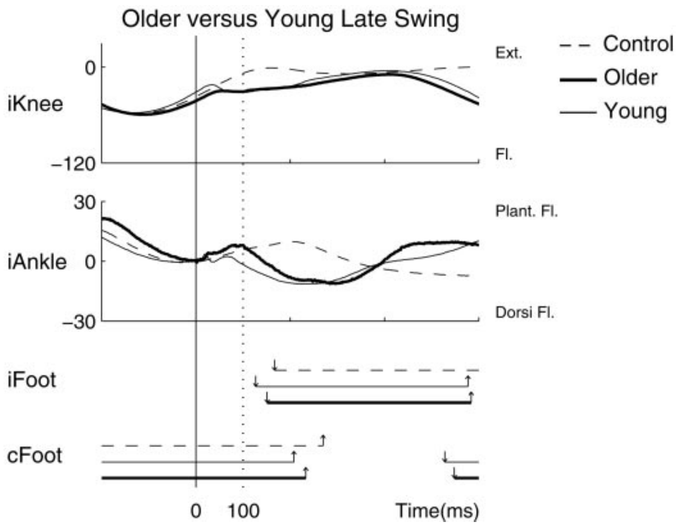

FIG. 9. Pooled joint angle changes (Degrees; not subtracted) during the late swing lowering strategy for the ipsilateral knee (iKnee) and ankle (iAnkle) of all older and all young adults. Angle at standing position is zero. Two *bottom traces* show stance phases of the ipsilateral (iFoot) and contralateral foot (cFoot). Same format is used as in Fig. 8.

iBF for deceleration of the forward sway, the lower responses in these muscles (although not significant in iRF) might be responsible for the somewhat later placing of the foot on the treadmill in older adults. In addition, there was no consistent difference between young and older adults in the amount of shortening of the corresponding contralateral stance phase (older 47 ms, SD 19 ms, about 7%; young 62 ms, SD 29 ms, about 9%; Wilcoxon rank-sum test, $P = 0.30$).

DISCUSSION

The present study reports on stumbling responses in the elderly. The responses were compared with stumbling responses of younger adults (as published in Schillings et al. 2000). Changes in the stumbling reactions of older adults compared with those of young adults might elucidate factors that contribute to the increased risk of falling in the elderly. The results showed that stumbling over obstacles induces similar behavioral responses ("strategies") in elderly as in young adults. Early-swing perturbations led to elevating strategies (foot directly lifted over the obstacle after contact with the obstacle), whereas late-swing perturbations induced lowering strategies (foot quickly placed on the treadmill and lifted over the obstacle in the succeeding swing). Similarly, the general pattern of muscle activation underlying these strategies was consistent with the pattern observed in young adults. However, some important differences in the stumbling responses of older and young adults occurred. First, elderly had less success in avoiding the obstacle after tripping in early swing because more older adults had a secondary contact with the obstacle. Second, after both early- and late-swing perturbations, the characteristics of the motor responses, such as response latency and amplitude, varied between both groups.

Early-swing elevating strategy

Response amplitudes of ipsilateral upper leg muscles (iBF and iRF) were smaller in older adults than those in young adults during the early-swing elevating strategy. Most markedly decreased were the RP2 and RP3 response amplitudes. The decreased muscle responses were associated with a reduction of the lengthening of the ipsilateral swing phase during obstacle crossing. Shorter swing durations might result in smaller safety margins during stepping over the obstacles, leading to an increased risk of falling in older adults. The video observations clearly supported this interpretation, showing more secondary contact with the obstacle in older adults. In these cases the ipsilateral foot was properly elevated above the obstacle but, instead of placing the foot over the obstacle, it was placed on top.

The video observations revealed that older adults mostly made shorter steps (in distance) during stumbling than during unperturbed walking, whereas young adults mostly increased their step length. This finding is in agreement with studies on obstacle anticipation on a walkway (Begg 2000; Chen et al. 1991; McFadyen and Prince 2002). In these studies, older adults used shorter stride lengths and reduced toe clearances during stepping over obstacles, compared with those of young adults. This can be interpreted as a preference of the elderly to use smaller steps over the obstacle, possibly to reduce the effort and keep their gait more consistent with their normal cadence of walking.

However, a large disadvantage of shortening the step is the increased risk of falling from a trip because the center of mass presumably tends to be anterior to the position of foot contact with the ground at the time of landing. This makes it difficult to slow the forward body rotation imposed by the perturbation, which was described earlier for postural perturbations during stance (i.e., treadmill accelerations; Owings et al. 2001). In the present study, it is unknown whether the center of pressure relative to the base of support was indeed different for older and young adults because these parameters were not measured.

Knee flexion during the elevation of the ipsilateral foot over the obstacle also tended to be reduced in older adults compared to that in young adults. Although this difference was not significant with the current number of subjects, this finding may be an important factor in studies on falling in the elderly because less knee flexion could result in lower clearance of the foot over the obstacle with smaller safety margins. This increases the risk that the foot reencounters the obstacle, which induces a second obstruction of the foot.

In contrast to the ipsilateral responses, the amplitudes of the upper leg muscle BF in the contralateral leg (RP4) were larger in older adults than those in young adults after early-swing perturbations. These BF responses on the contralateral side are considered to be of utmost importance in restoring balance. It has been suggested that in response to perturbations of the swinging leg, BF responses in the supporting leg could contribute to the stability of the upper body (Dietz et al. 1986; Eng et al. 1994). During the elevating strategy, the forward rotation of the body and the trunk needs to be stopped to regain balance (Eng et al. 1994; Grabiner et al. 1993; Pavol et al. 2001). Pijnappels et al. (2002, 2004a,b) described that a push-off force generated by hamstring and gastrocnemius muscles in the support limb contributes to the reduction of the forward rotation. Thus the increased cBF activity of the older adults in the present study might reflect an increased effort of these subjects to stabilize the upper body in early swing, which possibly points to a higher demand for trunk stabilization.

Late-swing lowering strategy

In late swing, older adults showed significantly lower RP3 responses in some muscles (iBF and iSO) compared with those in young adults. These muscles can contribute to an early foot placement because iBF contributes to slowing down the forward swing and iSO (and iRF) can take up body support in preparation of the early foot placement. The smaller muscle activity in older adults results in a somewhat later placement of the ipsilateral foot compared with that in the young, although the relative delay (25 ms) was not significant. In studies on stumbling reactions after tripping on a walkway it has been described that a delayed foot placement during the lowering strategy is associated with falling in older adults (Pavol et al. 2001; Van den Bogert et al. 2002). Pavol et al. (2001) showed that older adults who fall quickly after placing their foot during the lowering strategy had a latency of touchdown, which was about 100 ms longer than the latency of those who did not fall. Delayed foot placement leads to increased forward rotation of the body and to the center of mass being in front of the base of support during the limb loading of the ipsilateral leg. Thus the support force of the lowered limb cannot be used to stop the trip-induced forward body rotation (Pavol et al. 2001; Van den Bogert et al. 2002).

Underlying mechanisms of reduced ipsilateral RP2–RP3 responses in the elderly

As mentioned earlier, the amplitudes of the RP2 and RP3 responses during stumbling were mostly smaller in older adults than in young adults. A reduction of medium- and long-latency muscle response amplitudes in the elderly has also been described in previous studies with other types of unexpected perturbations during walking or stance. For example, in studies in which subjects slipped during walking, Tang and Woollacott (1998) found a reduction in muscle response amplitudes in both the ipsilateral and contralateral leg muscles of older adults. Some studies, in which stance was perturbed by sudden platform rotations in various directions, reported that the balance-correcting responses in TA and SO showed smaller amplitudes in the elderly than those in young adults in the interval between 120 and 220 ms after perturbation (Allum et al. 2002). However, others did not find major age-related changes in the amplitude of the responses (Nardone et al. 1995).

The underlying mechanism of the decreased response amplitudes during the stumbling reactions in aging could be explained by both peripheral and central changes. It was shown that the sensitivity in the proprioceptive sensory systems decreases with aging (Alexander 1994; Horak et al. 1989; Nadler et al. 2002; Stelmach and Worringham 1985; Verrilo 1980). In addition, the number of motoneurons is reduced (Mynark and Koceja 2001). Further, it is well known that during aging, the muscle properties change, resulting in muscle atrophy and reduced numbers of motor units, with especially a decrease of type II muscle fibers (Bouche et al. 1993; Maki and Fernie 1996; Merletti et al. 2002; Mynark and Koceja 2001).

On the other hand, changes in the CNS could also contribute to the reduced response amplitudes in older adults. Medium-latency reflex responses are considered to travel through a short train of spinal interneurons (Dietz 1992; Jankowska 1992; Nardone and Schieppati 1998). It has been suggested that age-related decreases in the supraspinal facilitatory drive on these interneuronal circuits interposed in these reflexes (Dietz 1992) might play a role. In addition, the excitability of aging motoneurons could be reduced by increased levels of presynaptic inhibition with aging (Koceja and Mynark 2000; Morita et al. 1995).

Latency of medium-latency responses

The present finding that older adults show significantly longer response latencies (prolongation 10–19 ms) in some muscles such as iBF, iTA, iSO, and cBF is in agreement with previous studies in which another type of unexpected perturbation was introduced during stance or walking (Nardone et al. 1995; Stelmach et al. 1989; Tang and Woollacott 1998, 1999). For example, in studies on slips during walking in early stance, the response latencies in the ipsilateral muscles of the anterior side of the leg (TA and RF) of the elderly increased by about 20 to 50 ms (see Fig. 2 in Tang and Woollacott 1998), whereas the ipsilateral muscles of the posterior side of the leg were not significantly different. In contrast, in the present stumbling experiments, we have found that muscles on both the anterior (iTA) and the posterior side (iBF and iSO) showed significantly delayed latencies.

Medium-latency reflex responses are most likely attributable to proprioceptive group II afferents (Dietz 1992; Jankowska 1992; Nardone et al. 1995, 1998; Schillings et al. 2000) or cutaneous afferents (Schillings et al. 2000). For cutaneous afferents, the latency increase could be accounted for by changes in the sensory thresholds of these afferents (Baloh et al. 2003; Verrillo 1980). Alternatively, there might be a decrease in nerve conduction velocity of the afferents, as has been suggested for the group II afferents by Nardone et al. (1995). Moreover, the latency increase could be a result of changes in the CNS (Tanosaki et al. 1999; Tobimatsu et al. 1998) or the alpha-motoneurons (Mynark and Koceja 2001).

RP1 short-latency responses

During the stumbling reactions, older adults showed significant short-latency responses (latency ≅ 45 ms) in the muscles of the ipsilateral leg after the collision with the obstacle. The latency of the short-latency response in iSO was 6 ms longer in older adults (this study) than that in younger subjects (Schillings et al. 1999). Similar lengthened latencies of short-latency stretch or H-reflexes in older adults have been described in studies on reflexes after stance perturbations (Nardone et al. 1995; Sabbahi and Sedgwick 1982; Scaglioni et al. 2002). The increase of these latencies in older adults can largely be attributed to a decrease of the peripheral nerve conduction velocity (for review see Mynark and Koceja 2001). However, slowing of central synaptic transmission at the spinal level might also play a role (Mynark and Koceja 2001; Sabbahi and Sedgwick 1982).

Regarding the amplitudes of short-latency responses, many authors described that for static body postures (sitting, prone, and standing) the amplitudes of either H-reflexes or tendon tap reflexes decline with normal aging (Angulo-Kinzler et al. 1998; Bouche et al. 1993; Kido and Stein 2002; Koceja et al. 1995; Mynark and Koceja 2001; Sabbahi and Sedgwick 1982; Scaglioni et al. 2002; Van Rey et al. 2002). Correspondingly, studies describing reflex responses during walking showed that in phases of the walking cycle, in which responses are prominent (in the stance phase), the amplitude of the soleus H-reflex was found to be smaller in elderly than that in young adults (Chalmers and Knutzen 2000; Kido and Stein 2002). However, during the swing phase of walking, H-reflexes were very small in both young and older adults and the 2 groups were not significantly different from each other (Chalmers and Knutzen 2000). This is similar to our results, which also showed no significant difference between the RP1 response amplitudes of older and young adults. It is suggested that under a dynamic condition, such as the swing phase of walking, the responses are too small and variable to allow a convincing identification of subtle reductions in amplitude.

In summary, the present results suggest that the increased risk of falling after a trip in the elderly results from changed muscle responses in the stumbling reactions of older adults. The older adults in the present study were all healthy community dwellers and had no history of falls. The age effects described might be even more pronounced if the group of elderly was at a more advanced age. In addition, in everyday life the circumstances of a trip could be more complex for older adults (such as circumstances with reduced vision, reduced attention, higher obstacles, and uneven floors). Further studies are needed to test these hypotheses.

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