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Bilateral early activity in the hip flexors associated with Falls in Stroke Survivors: preliminary evidence from laboratory-induced falls

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Abstract

Objective—Falls are the most common and expensive medical complication following stroke. Hypermetric reflexes have been suggested to impact post-stroke balance but no study has evaluated reflex amplitudes under real conditions of falls in this population. Our objective was to quantify the early reflexive responses during falls induced in the laboratory.

Methods—Sixteen stroke survivors were exposed to posteriorly directed treadmill perturbations that required a forward step to maintain a balance. Perturbations differed in terms of treadmill translation displacement, velocity, and acceleration. EMG amplitudes were compared between Fall/Recovery trials, as well as Fallers/Non-Fallers at two different time windows: 50–75 and 75–100ms.

Results—Sixteen of 86 trials resulted in falls by nine subjects (Fallers). While no differences were found between 50–75ms, EMG amplitude in the paretic rectus femoris muscle was larger between 75–100ms during Fall trials. Further, a bilateral increase in RF activity was seen in Fallers but not Non-Fallers. Interestingly, the bilateral increase was related to perturbation intensity (larger EMG activity with larger perturbations) in Fallers, but again not in Non-Fallers.

Conclusions—Heightened early recovery hip flexor activity between 75–100ms is associated with falls and Fallers post-stroke.

Significance—Though requiring replication and expanded subject pools, these preliminary results reflect a possible clinically meaningful relationship between heightened reflexive responses and fall risk. Future work should evaluate the underlying mechanisms driving these heightened

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Conflict of Interest

University of Illinois at Chicago owns a patent on some technology used in the ActiveStep treadmill system and consequently there is an institutional conflict of interest.

Mark D. Grabiner is an inventor of the ActiveStep system but has no conflicts of interest to declare with regard to the present study.

reflexes (e.g. stretch, startle) such that future rehabilitation techniques can address this abnormal response.

Keywords

balance; reflex; startle; stepping; trunk

1. Introduction

Stroke is the leading cause of disability in the US with an additional 800,000 incidents occurring each year (CDC, 2012). Falls present a major risk for stroke survivors, 40% of whom experience a serious fall within their first year (Persson et al., 2011). Up to 69% of falls by stroke survivors result in injuries. Despite the importance of the problem, there is surprisingly little information about what factors contribute to falls in stroke survivors. With few exceptions, the literature has focused on relating metrics of post-stroke static balance (where stepping did not occur) and impairment (clinical scores) to fall outcomes in the acute care setting or in the community (Divani et al., 2011; Ikai et al., 2003; Marigold et al., 2004; Marigold and Eng, 2006; Persson et al., 2011; Weerdesteyn et al., 2008). While informative, these studies provide little information about what specific impairments during a stroke-survivor's response to dynamic balance challenges lead to a fall.

Stroke survivors have known deficits in balance and reflexive control; however the specific attributes of a stroke survivor's balance and reflexive response that lead to a fall are unknown. Stroke survivors have poor automatic postural responses during small disturbances (those that do not result in a fall) including delayed and diminished muscle activation (Badke and Duncan, 1983; Diener et al., 1985, 1984; Dietz and Berger, 1984; Ikai et al., 2003; Marigold et al., 2004; Marigold and Eng, 2006) and abnormal muscle activation patterns (Badke and Duncan, 1983; Di Fabio, 1987). Further, the early reflexive responses that contribute to balance (e.g. stretch, startle) are abnormal (Finley et al., 2008; Honeycutt and Perreault, 2014, 2012; Jankelowitz and Colebatch, 2004; Trumbower et al., 2010). We have recently demonstrated that the inability to arrest and reverse trunk flexion and take an appropriate compensatory step were the most critical contributors to a failed recovery during balance disturbance (i.e. fall) in stroke survivors (Honeycutt et al., 2016). The extent to which a poor or abnormal early muscular response contributes to these deficits is unknown.

The objective of this study was to quantify the early postural response in lower extremity muscles during balance perturbations large enough to evoke a fall. Additionally, we will compare Fallers and Non-Fallers to determine if abnormalities in the early muscular response differ in Fallers even when compared to other stroke survivors. We exposed stroke survivors to posteriorly-directed perturbations that have been shown to mimic an overground trip (Owings et al. 2001). As stroke survivors have delayed, weaker muscle responses and diminished capacity to stabilize the trunk (Handley et al., 2009; Langhorne et al., 2009; Sridharan et al., 2009), we hypothesized that falls would be accompanied by smaller EMG amplitudes. Conversely, we found heightened early muscular responses, particularly in Fallers. We discuss our findings with respect to the different potential mechanisms

contributing this observed heightened reflexive response, as well as the clinical implications for fall prediction.

2. Methods

2.1. Subjects

Sixteen subjects with a unilateral brain lesion from a stroke participated in this study. Inclusion/exclusion criteria were the ability to stand unassisted for 5 minutes, no major vertebral or lower extremity surgery/injury in the past year, and no history of fainting. Subjects were recruited at the Rehabilitation Institute of Chicago (RIC) and Northwestern University. Experiments were conducted at the University of Illinois at Chicago (UIC). All methods were approved by both RIC/Northwestern and UIC Institutional Review Boards. Informed, written consent was obtained from all subjects.

2.2. Protocol

Subjects' weight, height, leg dominance prior to stroke, stroke type, stroke date, and affected limb were recorded. Balance and mobility were quantified using Berg balance, 5 times sit-to-stand, and 10m walk (comfortable/fast) tests. Questionnaires evaluating fall history, fear-of-falling, and physical activity (PASE: Physical activity Scale for the Elderly) were administered. Graded anteriorly- and posteriorly-directed perturbations intended to require a stepping response to avoid falling were delivered while subjects stood on a dual-belt, stepper motor driven and computer controlled treadmill (ActiveStep™, Simbex, Lebanon, NH). Subjects wore a ceiling-mounted upper body harness that was adjusted to allow natural falling in an unobstructed way. However, in the event that the subject was not able to establish dynamic stability following the perturbation, the harness safely and comfortably prevented the subjects' hands and knees from contacting the treadmill belts.

Subjects experienced 12 treadmill perturbations (6 posteriorly directed and 6 anteriorly directed) that required a stepping response to avoid a fall. Perturbation direction was defined with respect to the direction that the subject was facing, with anteriorly-directed perturbations employed to ensure the subject could not predict the next perturbation. Difficulty level (level 1, level 2, and level 3, i.e. low, moderate and high, respectively) was varied by modulating the displacement, velocity and acceleration (Table 1). The perturbation direction (anterior/posterior) was randomized, but the difficulty level was always delivered from low to high. Three subjects received only posteriorly-directed perturbations. However, because two of these three subjects fell and no other differences in their responses were observed, it was considered important to include the data from these subjects.

Trials were classified as either a "Fall" or a "Recovery". A fall was recorded when a subject became unambiguously supported by the harness. Subjects were classified as "Fallers" if they fell at least once during the experiment and "Non-Fallers" if they did not fall.

2.3. EMG recording and data analysis

Surface EMG was collected using bipolar Ag/AgCl electrodes (Noraxon Dual Electrodes, #272, Noraxon USA Inc., AZ) placed bilaterally over the rectus femoris (RF),

semitendinosus (ST) and gastrocnemius medialis (GM) muscles of both the paretic and non-paretic sides. EMG signals were amplified and conditioned using a band-pass filter of 10–1000 Hz. The resulting signals were anti-alias filtered using a 5th order Bessel filter with a 500 Hz cut-off frequency and sampled at 2500 Hz (PCI-DAS1602/16; Measurement Computing, MA). The signals from the first two subjects were sampled at 1200 Hz, but as this sampling rate is still sufficiently above the Nyquist limit, these subjects were not excluded from the analysis. Unprocessed EMG data were exported and further analyzed in MATLAB 2015b (The MathWorks, Inc., Natick, MA).

Digitized EMG signals were high-pass filtered at 30Hz using a 2nd order Butterworth filter, the mean background signal was removed, and the signal was then rectified and normalized. Mean background was calculated over a 500 ms long time interval preceding the onset of the perturbation. Normalization was based on the signal amplitude from the maximal level observed over all trials performed by each subject. Each EMG recording was evaluated visually to ensure accuracy and to exclude any trials during which the subject may have moved prior to the onset of perturbation.

To quantify reflex activity, EMG amplitudes were calculated as the area under the curve for 50–75 and 75–100ms following the perturbation onset. To ensure background activity was matched, background amplitudes were calculated as the area under the curve during a 60ms window preceding the perturbation onset and statistically compared across conditions (see below).

2.4. Kinematics

Twenty-two passive-reflective markers were placed over landmarks on the upper extremities, lower extremities and trunk using a modified Helen Hayes marker set (Kadaba et al., 1990). The three-dimensional locations of the markers were recorded at 120 Hz by an 8 camera motion capture system (Motion Analysis Co., Santa Rosa, CA) and filtered using a 4th order Butterworth with a 6 Hz cutoff frequency (Cortex 2.5.2, Motion Analysis Co., Santa Rosa, CA). A 13-segment rigid body model was constructed using the marker positions and kinematic variables were computed using custom software MATLAB 2015b (The MathWorks, Inc., Natick, MA).

In order to determine if alterations in EMG amplitude differences were related to differences in early body movement, we evaluated trunk flexion (sagittal plane angle of the line connecting the center of the pelvis to the midpoint of the line between the shoulder markers relative to initial starting position) at 50 and 75ms following perturbation onset.

2.5. Statistics

Because delayed and weak muscular responses are commonly observed in stroke survivors (Handley et al., 2009; Langhorne et al., 2009; Sridharan et al., 2009), we hypothesized that falls would be accompanied by smaller EMG amplitudes. To test this hypothesis, we compared reflex amplitudes between (a) Fall and Recovery trials, (b) Fallers and Non-Fallers, and (c) different perturbation levels. To ensure that any differences found in the muscle amplitudes were not related to differences in background activity or early changes in body movement, we also analyzed background muscle activity during a 60ms window

before perturbation onset (Capaday et al., 1994; Mrachacz-Kersting et al., 2004), and trunk flexion at 50 and 75ms after the onset of perturbation. All statistics were completed using R (R Core Team, 2016).

All dependent variables (EMG amplitudes, trunk flexion, and background EMG amplitude) were compared using a linear mixed-effect model ANOVA with either trial type (Fall or Recovery), subject type (Faller or Non-Faller), or perturbation level (1 - low, 2 - moderate or 3 - high) as the independent factors. The effect of perturbation level was compared separately for Fallers and Non-Fallers. Post-hoc analyses for perturbation levels were conducted using Tukey HSD. Separate analyses were also conducted for each of the recorded muscles. Between-group comparisons (Fallers vs. Non-Fallers) for subject characteristics and clinical scores were conducted using independent t-tests.

In all ANOVA analyses, subjects were treated as a random factor and equal variance was not assumed. We considered $P < 0.05$ to be statistically significant and $0.05 < P < 0.06$ as trending towards significance. All the results are reported with standard errors ($\pm SE$).

3. Results

86 trials from 16 subjects were included in the analyses, out of which level 1 perturbations involved 31 trials, level 2 – 28 trials, and level 3 – 27 trials. Out of 86 trials 16 resulted in falls, of which 13 occurred following a level 3 perturbation, the largest posteriorly-directed perturbation. Consequently, Fall vs. Recovery differences are reported only for level 3. Two subjects fell following the smallest perturbation (level 1), one of whom fell twice and did not initiate a recovery step. For safety, the investigator decided not to have this subject attempt level 2 and 3 perturbations.

Nine subjects were classified as Fallers and six as Non-Fallers. No differences were found between Fallers and Non-Fallers with regard to the age, BMI, type and duration of stroke, and performance on clinical tests (all $P > 0.05$; Table 2).

Falls were preceded by increased EMG amplitude in the pRF muscle during the 75–100ms time window (Fig. 1). Between-group differences for all muscles during the 50–75ms time window were not different (pRF: $F_{1,11} = 1.65$; $P = 0.22$ / pST: $F_{1,9} = 1.3$; $P = 0.28$ / pGM: $F_{1,10} = 0.69$; $P = 0.42$ / npRF: $F_{1,11} = 0.81$; $P = 0.38$ / npST: $F_{1,11} = 0.3$; $P = 0.59$ / npGM: $F_{1,12} = 0.27$; $P = 0.60$). However, the pRF muscle showed early onset EMG activity of less than 100ms during Fall trials (Fig 1B). Indeed, only 25% of Recovery trials generated a quantifiable pRF onset less than 100ms whereas 69% of Fall trials generated onset latencies less than 100ms. While this response appears to be bilateral with npRF showing onset latencies less than 100ms in Fall trials during 64% of trials compared to only 14% of Recovery trials, only pRF showed statistically larger EMG activity during the 75–100ms time window (Fig. 1A: pRF: $F_{1,11} = 7.4$; $P = 0.019$). Between-group differences between all other muscles during the 75–100ms time window were not significant (pST: $F_{1,9} = 1.4$; $P = 0.26$ / pGM: $F_{1,10} = 0.49$; $P = 0.49$ / npRF: $F_{1,11} = 0.95$; $P = 0.34$ / npST: $F_{1,11} = 0.13$; $P = 0.71$ / npGM: $F_{1,12} = 0.16$; $P = 0.69$).

While no differences existed between Fallers and Non-Fallers during the 50–75ms time window, Fallers showed bilaterally larger RF activity during the 75–100ms time window (Fig. 2). Both pRF and npRF showed larger EMG amplitudes during the 75–100ms window (pRF: $F_{1,11}=7.6$; $P = 0.018$ and npRF: $F_{1,11}=5.4$; $P = 0.039$) during level 3 perturbations. No differences were found on the smaller perturbations (Level 1 & 2) or during the 50–75ms time window (all $P>0.05$).

Fallers, but not Non-Fallers, demonstrated bilateral, perturbation level-dependent behavior (increasing EMG amplitude with more intense perturbation) during the 75–100ms time window (Fig. 3) in RF. Both pRF and npRF showed increasingly larger EMG activity with perturbation level during the 75–100ms time window in Fallers (Fig. 3A) but not Non-Fallers (Fig. 3B). In Fallers, there was an increase in pRF amplitude from level 1 to 2 ($F_{2,20}=7.28$; $P = 0.013$) and level 1 to 3 ($F_{2,23}=11.56$; $P=0.0025$). Similarly, there was an increase from level 1 to 3 ($F_{2,17}=16.7$; $P = 0.0006$) and level 2 to 3 ($F_{2,19}=13.59$; $P = 0.0018$) in npRF in Fallers. No perturbation-level differences were found for Non-Fallers during the 75–100ms time window (all $P>0.05$). No perturbation-level differences were found for either Fallers or Non-Fallers during the 50–75ms time window.

Background activity (muscle activity prior to perturbation) was not different for any of muscles and conditions (all $P>0.05$), -suggesting that the differences observed between Fall vs. Recovery trials, Fallers vs. Non-Fallers, and perturbation levels were not caused by the heightened activity in the spinal network.

No differences were found in trunk kinematics at 50 or 75ms after perturbation onset between Fall/Recovery trials or Fallers/Non-Fallers. Though pRF activity was larger during Fall trials (see above), no difference in trunk flexion was seen between Fall vs Recovery trials at 50 ($F_{1,12}=2.94$; $P=0.11$) or 75ms ($F_{1,12}=1.20$; $P=0.29$). Though bilateral increases in RF activity were seen in Fallers on level 3 perturbations, no difference in trunk flexion was seen at 50 ($F_{1,12}=0.90$; $P=0.36$) or 75ms ($F_{1,12}=0.18$; $P=0.67$).

4. Discussion

4.1. Summary

The objective of this study was to quantify the early postural response in lower extremity muscles during balance perturbations large enough to evoke a fall in individuals who have had a stroke. Given previous evidence of weakened and poor trunk stability (Badke and Duncan, 1983; Diener et al., 1985, 1984; Dietz and Berger, 1984; Ikai et al., 2003; Marigold et al., 2004; Marigold and Eng, 2006), we hypothesized that falls would be accompanied by significantly smaller early muscular response amplitudes. Conversely, we found that pRF had larger reflexive activity during Falls. Furthermore, we observed a bilateral increase in RF muscle amplitude in Fallers. This increase was perturbation-level dependent in Fallers while Non-Fallers did not demonstrate an increase in RF activity or scaling effects across perturbation levels. The early activity was mid-to-long long latency occurring between 75–100ms after the perturbation onset. No differences were found in the 50–75ms time window. Background muscle activity and trunk flexion kinematics were matched suggesting a pathophysiological cause and not an artifact of increased excitation of the spinal motor

neuron network or differences in early body movement prior to a larger perturbation. Though a larger dataset (including additional subjects and muscles) and replication is needed, we interpret these preliminary results as suggesting a potentially clinically meaningful relationship between heightened reflex responses and fall risk.

We know of only 2 other groups that have evaluated actual falls in the laboratory by stroke survivors (Kajrolkar et al., 2014; Marigold and Eng, 2006; Salot et al., 2016). We distinguish the present work from that of previous authors from these by having evaluated forward stepping perturbations – those that more closely resemble a trip (Owings et al. 2001). Recent evidence suggests that the neural mechanisms governing the response to slips (like those evaluated above) is distinctive from that during trips (Nonnekes et al., 2013). Given that trips are a leading cause of falls during gait (Berg et al. 1997), it is critical that we evaluate the muscular response during forward stepping responses as a potential contributing mechanism underlying falls.

4.2 Neural mechanisms

As this was a preliminary study that requires replication and increased subject numbers, we cannot fully elucidate the neural mechanisms underlying the observed heightened activity. However, these data do give certain insights. First, the response is mid-to-long latency, occurring after 75ms. This likely rules out short-latency stretch reflexes or spasticity as a major underlying cause. Attempts to relate spasticity to falls has been mixed in the past (Dietz and Sinkjaer, 2007; Nardone et al., 2001; Sommerfeld et al., 2004). Our work further adds to the work that has suggested that at least a short-latency stretch response is not linked to falls. Though mid-to-long latency stretch reflex responses play a significant role in spasticity of the upper extremity, their contributions to lower extremity responses are less robust (Finley et al., 2008).

Bilateral responses further argue against a spasticity-driven explanation. Rather, they point towards other bilateral reflexes such as startle. Fallers showed a bilateral increase in RF activity compared to Non-Fallers. Further, a perturbation-level dependent sensitivity of the RF muscle response was also found bilaterally in Fallers but not Non-Fallers. While amplitude differences were not statistically significant, we also found that npRF showed onset latencies less than 100ms in 64% of Fall trials compared to only 14% of Recovery trials. The startle response, which has been shown to be abnormally heightened in stroke survivors (Honeycutt and Perreault, 2014, 2012; Jankelowitz and Colebatch, 2004), is bilateral in nature and has an average onset latency of 75ms (Rothwell, 2006; Valls-Solé et al., 1999), which corresponds to our results showing increased responses during the 75–100ms time window. This hypermetric startle reflex has been shown to interfere with upper extremity postural adjustments (Honeycutt and Perreault, 2014, 2012). Though the functional role of the startle reflex during whole-body postural adjustments remains unclear, it is certain that startle is activated during perturbations (Blouin et al., 2006; Nonnekes et al., 2013) and, further, that it is linked to heightened activity during the first trial (Siegmund et al., 2008). Sawers et. al. (2017) also proposed startle as the mechanism underlying the unique “all on” muscle synergy observed during slip conditions in older adults. They observed that Fallers had strong bilateral co-activation patterns. While their observation was

in older adults, it is remarkably similar to our observations in stroke survivors. Thus, when our present work is viewed in light of recent work in the literature, the heightened hip flexor activity seen during Fall trials and Fallers could result from a hypermetric startle reflex.

4.3. Functional consequences

Large trunk flexion during a recovery step has been shown to be a determinant of anteriorly-directed falls by stroke survivors, healthy middle age and older women, lower extremity amputees, and middle age and older women with knee osteoarthritis (Crenshaw et al. 2013; Crenshaw et al. 2012; Honeycutt et al. 2016; Owings et al. 2001; Pater et al. 2016; Rosenblatt & Grabiner 2012). We have previously shown that stroke survivors who are Fallers have larger trunk flexion and trunk flexion velocities at step initiation and step completion (Honeycutt et al., 2016). Given that the pRF muscle is a hip flexor that influences sagittal plane pelvic, and potentially trunk orientation, it is possible that a larger RF muscle activation could contribute to larger trunk flexion in Fall trials. However, the extent to which this relationship may be causal has not been established for post-stroke patients. If a biomechanically causal relationship can be confirmed, then an important question relates to whether the relationship is clinically modifiable.

Early reflexive responses are generally considered to have a small functional effect on motion, in part, because of their small amplitude compared to voluntary muscle activity. However, in our study we evaluated muscular responses only in 3 muscles. Therefore, other muscles may exhibit heightened responses as well, thereby increasing the aggregate effect.

4.4 Future directions

The present work represents a preliminary study and, as such, in the light of the potential clinical implications of the finding, further systematic study to replicate and extend the results seems prudent. Further, because the underlying neural mechanisms of the muscular responses have not been established further systematic study should 1) consider a larger muscle set including tibialis anterior and startle indicators such as the sternocleidomastoid, 2) evaluate the impact of differential loading between the paretic and non-paretic limbs, 3) examine co-activation, and 4) determine the presence of mid-to-long latency stretch and startle components.

4.5 Conclusions

By employing large, treadmill-delivered postural perturbations that serve as a surrogate for an overground trip, the present study established an association between hypermetric reflexive activity of multiarticular lower extremity muscles and falls by stroke survivors. This is the first study to evaluate the early reflexive response during perturbations large enough to evoke a fall in stroke survivors. While more research is needed to confirm this result and elucidate the mechanisms, the present findings offers the potential of using EMG measurements of reflexive activity to establish fall risk as well as a metric for the effectiveness of fall-prevention interventions for post-stroke rehabilitation.

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Abbreviations

pRF/npRF paretic/non-paretic rectus femoris

pST/npST paretic/non-paretic semitendinosus

pGM/npGM paretic/non-paretic gastrocnemius medialis

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Highlights

1. Stroke survivors have larger early activity in the paretic hip flexors when a Fall occurs.
2. Fallers, but not Non-Fallers, demonstrated bilaterally larger hip flexor activity.
3. Early onset of hip flexor activity may have a clinically meaningful relationship to falls.

Figure 1. EMG amplitude comparisons for Fall vs Recovery trials

A) Group results of the mean EMG amplitude for 75–100ms during level 3 perturbations are depicted. No differences were found during the 50–75ms time window for all muscles evaluated (RF, ST, GM). Only pRF showed a statistically significant difference between Fall and Recovery trials during the 75–100ms time window. B) Examples of pRF activity during Fall trials (top) and Recovery trials (bottom) are depicted. Light and dark gray boxes show the 50–75ms and 75–100ms time windows respectively.

Figure 2. EMG amplitude comparisons for Faller vs Non-Fallers by perturbation-level
 EMG amplitude is compared between Fallers and Non-Fallers during the 75–100ms time window. While no differences were found between levels 1 & 2, Fallers show increased EMG amplitudes bilaterally at level 3.

Figure 3. EMG amplitudes across perturbation levels between Fallers (top) and Non-Fallers (bottom)

While no perturbation-level differences were found in EMG amplitudes of Non-Fallers, the RF muscle showed a bilateral increase with perturbation level in Fallers.

Table 1

Characteristics of the different levels of perturbation.

	Distance (cm)	Peak velocity (m/s)	Peak acceleration (m/s ²)
Level 1	21.84	0.26	6.53
Level 2	29.44	0.64	15.88
Level 3	76.00	1.30	12.90

Table 2

Participant characteristics for Faller and Non-Faller groups.

	Fallers (<i>n</i> =9), mean (SE) or <i>n</i>	Non-Fallers (<i>n</i> =6), mean (SE) or <i>n</i>	P-value
Gender (M/F)	6/3	6/0	-
Age (year)	62.3 (1.19)	59.0 (0.93)	0.52
BMI (kg/m ²)	29.5 (0.42)	28.0 (1.05)	0.59
Hemiparetic side (R/L)	7/2	4/2	-
Dominant leg before stroke (R/L/unknown)	5/3/1	5/0/1	-
Type of stroke (ischemic/hemorrhagic)	5/4	5/1	-
Stroke duration (year)	10.2 (0.63)	7.0 (0.54)	0.26
Berg balance, max. 56	47.4 (0.80)	52.5 (0.44)	0.15
5 time sit to stand test (s)	22.2 (1.03)	23.1 (1.89)	0.88
10 m walk test, comfortable pace (s)	10.1 (0.96)	7.2 (0.32)	0.46
10 m walk test, fast pace (s)	9.2 (1.32)	5.3 (0.15)	0.47
Physical activity scale for the elderly (PASE)	137.5 (6.11)	167.1 (12.11)	0.42
Fear of falling, max. 64	30.9 (0.92)	24.5 (1.02)	0.16