**Increased use of stepping strategy in response to medio-lateral perturbations in the elderly relates to altered reactive tibialis anterior activity**

**Authors:** Maarten Afschrift[1], Robert van Deursen[2], Friedl De Groote[1], Ilse Jonkers[1]

**Author affiliation**

[1]Human movement sciences, KU Leuven, Belgium

[2]School of Healthcare Sciences, Cardiff University, UK

**Address for Correspondence**

Corresponding author: Maarten Afschrift

Tervuursevest 101 - bus 1501, 3001 Leuven, Belgium

maarten.afschrift@kuleuven.be

+32 16 37 64 80

Journal: **Gait & posture => 3000 words + 300 abstract (2949/3000 words)**

**Keywords**: Posture, aging, gait stability, simulation, balance

**Author Contributions**

MA RVD designed the experimental protocol, MA performed the experimental data collection, MA FDG implemented the simulation model, MA FDG IJ RVD contributed to data analysis and interpretation. All authors read, reviewed and approved the final manuscript.

**Acknowledgments**

Maarten Afschrift is a research fellow of the Flemish agency for scientific research (FWO-Vlaanderen). We thank Kate Jones for her help in the data collection.

**Supplementary material**

See OnlineSupplement.pdf

**Conflict of interest**

All authors hereby declare that there are no conflicts of interest.

**Competing interests**

The author(s) declare no competing interests.

**Abstract (298 / 300 words)**

*Background:* The influence of aging on reactive control of balance during walking has been mainly investigated in the sagittal plane, whereas balance control in response to frontal plane perturbations is largely unexplored in the elderly. This is remarkable, given that walking mainly requires active control in the frontal plane. An extensive gait perturbation protocol was used to test whether reactive control of walking balance changes with aging and whether these changes are more pronounced in the frontal than in the sagittal plane.

*Research question:* We hypothesize that alterations in reactive muscle activity cause an age-related shift from lateral ankle to stepping strategy in response to perturbations in the frontal and sagittal plane, and that the alterations in the frontal plane will be larger than the alterations in the sagittal plane.

*Method:* A treadmill-based perturbation protocol imposed frontal and sagittal plane perturbations of different magnitudes during different phases of the gait cycle. Motion capture and electromyography measured the response to the different perturbations in a group of eighteen young and ten older adults.

*Results:* Only for a small subset of the perturbations, reactive muscle activity and kinematic strategies differed between young and older subjects. When perturbation magnitude increased, the older adults relied more on a stepping strategy for inward directed frontal plane perturbations and for sagittal plane perturbation just before heelstrike. Tibialis anterior activity increased less in the older compared to the young subjects. Using simulations, we related tibialis anterior activity to backward and outward movement of the center of pressure in the stance foot and confirmed its contribution to the ankle strategy. We concluded that deficient tibialis anterior activity predisposes elderly to use stepping rather than lateral ankle strategies to control balance.

*Significance*: Rehabilitation targets for fall prevention in elderly need to also focus on ankle muscle reactivity.

## Introduction (637)

Falling is a major threat to the steadily growing population of elderly. Elderly are more at risk of falling, especially during walking [1]. Reactive control of balance in response to a perturbation relies on adjustment in muscle activity and joint kinematics to return to a steady-state walking pattern after perturbation [2–4]. Knowledge of the age-related differences in reactive muscle activity and kinematics in response to perturbations might help to understand the deficits in the control of balance during walking in elderly and in defining therapeutic targets to prevent falls.

The two main mechanisms for the control of balance during walking are (1) adjustment of the position of the center of pressure (COP) [5] and (2) adjustment of the magnitude and direction of the ground reaction force [6]. This study mainly focused on the control of the COP, through a stepping or ankle strategy [5]. Both strategies adjust the position of the COP with respect to the position of the extrapolated center of mass (Xcom) [7]. When using a stepping strategy, the COP location is controlled by adjusting the medio-lateral and anterior-posterior foot placement [2,3,7,8]. In the frontal plane, gluteus medius activity has been related to medio-lateral foot placement and is therefore considered to be the main contributor to the stepping strategy [2–4]. When using an ankle strategy, the COP location in the stance foot is adjusted by generating a torque around the ankle joint through reactive tibialis anterior and peroneus activity in the frontal plane (i.e. lateral ankle strategy [5,9]) and reactive gastrocnemius, soleus and tibialis anterior activity in the sagittal plane [10]. The corrective effect of the ankle strategy is smaller compared to the stepping strategy, because the movement of the COP is constrained by the border of the foot, especially in the medio-lateral direction.

The influence of aging on reactive balance control during walking has mainly been investigated with sagittal plane perturbations [8,11–13]. Older subjects need more steps to return to a stable gait pattern and have an increased COM movement after sagittal plane perturbations [8,11]. However, the influence of aging on walking balance control in the frontal plane is largely unexplored. This despite the fact that walking is mainly unstable in the frontal plane and therefore requires more active frontal than sagittal plane control [14–16]. The important role of the tibialis anterior and gluteus medius in the lateral ankle and stepping strategy suggests that altered function of these muscles could cause deficits in frontal plane balance control in the elderly. During perturbed standing, reduced tibialis anterior strength has already been related to a shift from ankle towards stepping strategy in the elderly [17,18]. In addition, reduced COP modulation, caused by deficits in ankle muscle activity, might elicit increased foot placement adjustments [19]. Other experimental studies of standing balance control suggest that hip abductor (i.e. gluteus medius) strength limits postural control performance in response to medio-lateral perturbations in the elderly [20]. In addition, decreased hip abductor strength was found to be a predictor of future fall risk [21] and to be larger in fallers as compared to non-fallers [22].

Currently, it remains unclear how aging affects balance control during walking, and if altered reactive activity is related to deficits in the control of balance during walking. We hypothesize that, comparable to standing balance control, elderly shift earlier from a lateral ankle to stepping strategy in reactive control of balance during walking both in the frontal and sagittal plane [17,18]. In addition, we hypothesize that age-related changes in frontal plane balance control are more pronounced than age-related changes in sagittal plane balance control and can be related to concurrent alterations in reactive tibialis anterior or gluteus medius activity. Therefore, gait was perturbed in a group of young and elderly subjects in medio-lateral and anterior-posterior direction during different phases of the gait cycle.

**Methods (772)**

Kinematics and muscle activity in response to medio-lateral or anterior-posterior perturbations with onset during different phases of the gait cycle were measured. 18 healthy young (age 21 ± 2 years, mass 70.7 ± 11.1 kg, height 1.78 ± 0.09 m) and 10 older adults (age 71 ± 4 years, mass 71.2 ± 11.8 kg, height 1.77 ± 0.11 m) without a history of falling, participated in the study. The study was approved by the ethics committee of the School of Healthcare Sciences, Cardiff University.

Perturbation protocol

After accommodating to split-belt treadmill (Grail, MotekForceLink) walking at a speed of 1.1 m/s for two minutes, subjects were exposed to three perturbation sessions separated by five minutes rest. A perturbation session consisted of 32 unique perturbations imposing two different perturbation magnitudes in four directions applied at four different instants in the gait cycle (Figure 1). Medio-lateral perturbations were induced by a sudden platform translation to the left or right and anterior-posterior perturbations were induced by a sudden increase or decrease in both belt speeds. The direction of loss of stability defined the perturbation direction, i.e. an increased belt speed caused the subject to fall forward and is considered a forward perturbation. All perturbations were applied immediately after left heelstrike (7.5% gait cycle, first double support), during early midstance (22.5% gait cycle), late midstance (37.5% gait cycle) or push-off (52.5% gait cycle, second double support) (see online supplement).

Measurements and data analysis

Kinematics in response to the perturbations were recorded at 200Hz by 12 Vicon T20 cameras (Vicon, Oxford) measuring the trajectories of 48 markers of an extended plug-in gait marker protocol. Ground reaction forces were collected with a sampling frequency of 1000Hz (MotekForceLink). Ground reaction forces and 3D marker coordinates were filtered using a 4th order low-pass Butterworth filter (cutoff frequency at 8Hz). Left and right gluteus medius, gastrocnemius, soleus and tibialis anterior activity was measured using surface electromyography (EMG) (Bortec Octopus EMG) at 1000Hz. EMG data were filtered with a second order IIR notch filter and a 4th order recursive Butterworth band pass filter (cutoff frequencies at 20-400Hz). Subsequently, the filtered signals were rectified and a linear envelope was created with a 4th order recursive Butterworth low-pass filter (cutoff frequency at 20 Hz).

Left and right heelstrikes were determined from the ground reaction forces. The muscle response to a perturbation was determined by subtracting the muscle activity during unperturbed walking from the activity after perturbation. Processed EMG data was averaged over the last 60 gait cycles of unperturbed walking, and processed EMG data of the gait cycle after the perturbation were interpolated to 1000 data points. EMG was normalized the maximal value measured during unperturbed walking. The muscle response was then quantified as the time integral of the difference in muscle activity during the first 300ms after perturbation onset.

A musculoskeletal model with 23 degrees of freedom was scaled to the subject's anthropometry and was used to calculate foot and COM kinematics from the recorded marker trajectories in OpenSim [23,24]. Stride width was computed as the average frontal plane distance between the left and right ankle joint center during double support. COM kinematics was computed in OpenSim using the BodyKinematics Analysis (i.e. kinematic method). The position of the Xcom was computed based on the position and velocity of the COM relative to the position of the ankle joint center at heelstrike [25]. The movement of the COP was computed as the average distance between the COP and the ankle joint center 150 to 300ms after the perturbation (i.e. first 150ms removed due to noise caused by inertia of the moving force plates, see limitations).

Symmetry between the left (i.e. inward) and right (i.e. outward) leg was assumed to evaluate the medio-lateral platform translations (Figure 1).

Outcome measures and statistical analysis

A generalized linear mixed effect model with variable intercept, bonferroni correction and a two-sided alpha level of 0.05 was used to evaluate the effect of the different perturbations and age groups on stride width, stride length and margin of stability and on the reactive muscle activity.

Secondary analysis using model-based simulation

For muscles with a different response in young and older adults, we evaluated the effect of reactive muscle activity on COP movement using our recently developed forward simulation workflow [26]. This allows evaluating the effect of an individual muscle response on the kinematics, which is impossible in the experimental approach where reactive activity of multiple muscles determine the COP movement. In addition, this method allows discriminating between the effect of the perturbation force (i.e. passive response) and reactive muscle activity (i.e. active response) on the movement of the COP (see [26] for details).

**Results (573)**

Measurements - Medio-lateral perturbations

In both age groups, timing of the medio-lateral perturbation significantly affected stride width, margin of stability, COP motion, and reactive muscle activity (for details Figure 2, 4). Large outward perturbations early in the gait cycle, increased stride width ($p<0.001$), decreased stride length ($p<0.001$), positioned the stance foot COP more medially [$p<0.01$] and Xcom more forward [$p<0.01$]. In addition, stance and swing leg gluteus medius activity increased while stance leg gastrocnemius activity decreased (Figure 4). Inward perturbations between 52 % and 77% of the gait cycle, decreased stride width (Figure 2b, $p<0.001$) and positioned the stance foot COP more laterally (Figure 5a, $p<0.001$) without changing stride length and Xcom position (Figure 2). Furthermore, stance leg soleus, gastrocnemius and tibialis anterior activity increased (Figure 4).

Despite the similarities in responses described above, differences in magnitude of reactive muscle activity and foot placement adjustments were observed in young and older adults following inward directed perturbations (i.e. 37.5 - 72.5% of the gait cycle). Older adults relied more on an inward stepping strategy, given the reduced stride width (Figure 2a, $p<0.01$). In addition, their stability decreased, given the decreased medio-lateral and anterior-posterior distance between Xcom and the foot at heelstrike for the inward perturbations at 37.5 and 52.5% of the gait cycle (Figure 2cd). Older adults increased reactive tibialis anterior activity less compared to the small perturbations (Figure 4a). Changes in foot placement, reactive tibialis anterior activity and COP movement were most pronounced for inward directed perturbations just before heelstrike (i.e. 37.5% of gait cycle, Figure 6).

Measurements - Anterior posterior perturbation

In both age groups, forward perturbations increased stride length ($p<0.001$, Figure 3) and increased stance leg soleus and gastrocnemius activity (Figure 4). Backward perturbations decreased stride length (Figure 3) and stance leg soleus activity (Figure 4), but increased stance leg tibialis anterior activity (Figure 4).

Despite the similarities in responses of young an older adults described above, differences in the magnitude of reactive muscle activity and adjustment of foot placement were observed in response to backward and forward perturbations at 37.5% of the gait cycle (i.e. perturbation onset just before heelstrike, Figure 3-4). Compared to young subjects, older adults reduced stride length more and increased forward Xcom position at heelstrike in response to the large backward perturbations at 37.5% of the gait cycle (figure 3b). For this perturbation, a larger increase in gluteus medius and gastrocnemius activity, but decrease in tibialis anterior activity of the outward leg (i.e. loading response leg) was observed (Figure 4c). In response to the forward perturbations at 37.5% of the gait cycle, older subjects increased stride length more than the young subjects (figure 3a). For this perturbation, a larger increase of tibialis anterior activity of the inward leg (i.e. push-off leg) was observed in the elderly compared to the young subjects (Figure 4b).

Simulation

To discriminate between the effect of perturbation force and the contribution of multiple muscles to the movement of the COP, we evaluated the isolated effect of reactive tibialis anterior, soleus and gastrocnemius activity on COP movement in simulation. This analysis was conducted for the inward directed perturbations and sagittal plane perturbations at 37.5% of the gait. Stance leg reactive tibialis anterior activity correlated significantly with the simulated lateral movement of the stance foot COP

($p<0.01$, $R^2=0.86$, Figure 5a). In addition, soleus and gastrocnemius activity shifted the COP movement forward, whereas tibialis anterior activity shifted the COP backward in the foot (Figure 5bc).

**Discussion (967)**

This study evaluated how aging influences the selection of ankle and stepping strategies in response to multidirectional perturbations of different amplitudes and in different phases of the gait cycle. When perturbation magnitude increased, older adults relied more on a stepping than an ankle strategy for sagittal plane perturbations applied just before heelstrike and for inward directed frontal plane perturbations (Figure 2-3). This shift in kinematic strategy is consistent with elderly shifting from ankle toward hip and stepping strategies at lower perturbation magnitudes of perturbed standing than young adults [17,18].

The experimentally observed decrease in reactive tibialis anterior activity may explain the observed shift from ankle toward stepping strategies in response to inward directed perturbations. Whereas, both young and older adults increased tibialis anterior activity in response to small inward directed perturbations, older adults failed to increase tibialis anterior activity (Figure 4) and the associated lateral movement of the COP with increasing perturbation magnitude (Figure 5) and adjusted foot placement, indicative of a stepping strategy (Figure 2). The observed relation between reactive tibialis anterior activity and COP movement (Figure 6) suggests that tibialis anterior activity controls COP movement [9].This was further confirmed using forward simulations that causally related the tibialis anterior activity to lateral COP movement (Figure 5). This suggests that reactive tibialis anterior activity is required to move the COP in the foot, and that a failure to generate timely and sufficient tibialis anterior activity necessitates COP position adjustment through a stepping strategy.

Similar as in standing balance control [17], altered tibialis anterior strength in the elderly may limit the capability to use a lateral ankle strategy in the elderly. Additionally, age-related changes in proprioception may decrease sensory acuity [27], which may induce poor detection of the perturbation and increased reaction time. Alternatively, the preference for the stepping strategy in the elderly might relate to a preference for a more robust control approach to cope with altered sensorimotor function or fear of falling. The stepping strategy is more robust against perturbations compared to the lateral ankle strategy as the potential COP excursion is larger in a stepping strategy compared to a lateral ankle strategy where it is constrained by the borders of the foot. Predictive simulations of perturbed walking, similar to our previous work in perturbed standing, may elucidate to what extent altered tibialis anterior function versus altered control goals (e.g. robustness versus efficiency) contributes to the premature change towards a stepping strategy in the elderly [18].

The increased use of a stepping strategy by elderly in response to forward and backward perturbations just before heel strike could not be explained by reduced COP modulation [19] due to altered tibialis anterior and gastrocnemius responses. This despite the fact that in simulation, reactive soleus and gastrocnemius activity contributed to forward COP movement (Figure 5) and tibialis anterior activity contributed to backward COP movement (Figure 5) [28]. Despite the increased gastrocnemius activity and decreased tibialis anterior activity in the older adults in response to a backward perturbation (figure 4), the measured COP movements in elderly were not different from young adults (Figure 5).

In contrast to our hypothesis, our results reject the hypothesis that deficits in gluteus medius function affect balance control in this group of healthy older adults. Indeed, kinematics of outward stepping, which is mainly controlled by reactive gluteus medius activity [7], was not different. In addition, the positive margin of stability, increased stride width and decreased stride length at the first step after perturbations shows that the outward stepping strategy was equally successful in both the young and older subjects.

Similar as in other perturbation experiments [9,29], we found that stance leg soleus and gastrocnemius activity is modulated following medio-lateral perturbations during walking (figure 4). This finding seems to confirm the simulation result that calf muscles contribute to the control of frontal plane angular momentum during walking by changing the ground reaction force magnitude and direction [6]. However, modulation of reactive calf muscle activity has also been interpreted as a mechanisms to compensate for reactive tibialis anterior activity (i.e. lateral ankle strategy) [9] or to control stability through push-off modulation [29].

Several limitations should be considered when interpreting the results. First, the perturbation magnitude was limited (Figure 1). Increasing perturbation magnitudes may reveal additional age-related changes in reactive control during walking. Second, despite its important role in the lateral ankle strategy [9], the reactive peroneus activity was not measured in this study. However, Hof et al. showed that the peroneus muscle mainly contributes to the lateral ankle strategy in response to outward directed perturbations [9], for which no kinematic differences between the young and elderly were found. Third, measured ground reaction forces and moments were corrected for the acceleration of the motion base. Nevertheless, these corrections might have influenced the COP location in the medio-lateral perturbations (Figure 5). Fourth, only the relation between reactive muscle activity and COP adjustments was studied, whereas reactive muscle activity might also contribute to balance control through adjustment of the ground reaction force magnitude and direction (i.e. inertial strategies). Likewise, no comprehensive analysis of the relation between reactive muscle activity and balance control mechanisms was formulated but only muscle responses discriminating young and older subjects were analyzed in simulation. Muscle responses not influenced by aging are therefore not discussed in this study, despite a potential major role in controlling balance. Finally, only healthy older adults without a fall history were included.

**Conclusion**

In conclusion, elderly presented a shift from ankle toward stepping strategies in response to inward directed perturbations and perturbations in the sagittal plane with onset before heelstrike during gait and this could be related to altered tibialis anterior, but not gluteus medius, reactive activity. Hence, our results suggest that gait perturbation paradigms for training of reactive recovery responses might benefit from an increased focus on this subset of perturbations.

**Conflict of interest**

All authors hereby declare that there are no conflicts of interest.

**Competing interests**

The author(s) declare no competing interests.

**Funding:**

Flemish agency for scientific research (FWO-Vlaanderen)

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**Figure captions**

Figure 1: **Overview perturbation protocol**. The perturbation was applied during different phases of the gait cycle (Pane A). When assuming symmetry between the left and right leg, the perturbation in the sagittal plane at 7.5% and 52.5% are similar (i.e. both perturbations start just after heelstrike of

the left and right leg respectively). In addition, when assuming symmetry, platform translations to the right at 7.5%, 22.5% and 37.5% of the gait cycle can be discussed as platform translations to the left at 57.5%, 72.5% and 87.5% of the gait cycle. We will refer to the right leg as outward leg and left leg as inward leg. A platform translation to the left causes a loss of stability to the right that is counteracted by either outward placement of the right foot hence termed the outward leg, or inward (cross-over) placement of the left foot, hence termed the inward leg (Pane B). Similarly, platform translations to the left during left stance are referred to as outward perturbations and platform translations to the left during right stance are referred to as inward perturbations (Pane A). The position, velocity and acceleration profile of the medio-lateral, forward and backward perturbations is shown in pane C.

Figure 2: **Change in stride length (a), stride width (b), sagittal margin of stability (c) and frontal margin of stability (d) at first and second heelstrike after perturbation compared to unperturbed walking.** The dot represents the average change in outcome variable and the vertical line the 95% confidence interval. The response to the large perturbation is shown in color (blue= young, red=elderly) and the response to the small perturbation is shown in gray. Responses significantly different from unperturbed walking are shown in bright colors for the large perturbation. Significant differences between the young and elderly for the large perturbations were highlighted with a black horizontal line. A small offset in onset timing of the perturbation was added to the data of the elderly for visualization purposes. Note that the first step is at heelstrike of the outward (i.e. right) leg for the perturbations with onset between 0%-40% of the gait cycle. The outward leg is in stance phase for perturbations with onset between 40-90% of the gait cycle, and the first step is therefore analyzed at heelstrike of the inward (i.e. left) leg.

Figure 3: **Change in stride length and sagittal MOS at first and second heelstrike after belt acceleration (a) and belt deceleration (b) compared to unperturbed walking.** The dot represents the average change in stride length and MOS and the vertical line the 95% confidence interval. The response to the large perturbation is shown in color (blue= young, red=elderly) and to the small perturbation is shown in gray. Responses significantly different from unperturbed walking are shown in bright colors. Significant differences between the young and elderly for the large perturbations are highlighted with a black horizontal line. A small offset in onset timing of the perturbation was added to the data of the elderly for visualization purposes (i.e. horizontal distance between red and blue data points). No changes in stride width and medio-lateral margin of stability were observed after perturbation compared to unperturbed walking.

Figure 4: **Reactive muscle activity in response to the perturbation (0-300ms).** The muscle response to a sudden medio-lateral platform translation (a), increase (b) and decrease (c) in belt speed is shown as a function of the onset timing of the perturbation during the gait cycle. The dot represents the average muscle response and the horizontal line the 95% confidence interval. The activity in response to the large perturbation is shown in color (young = blue, elderly = red) and to the small perturbation is shown in gray. Muscles responses significantly different from zero are shown in bright colors for the large perturbations. Significant differences between the young and elderly for the large perturbations were highlighted with a black square. The transition between stance phase of the inward and outward leg and the resulting transition between inward and outward perturbations during the gait cycle is shown on the top of pane a.

Figure 5: **Measured and simulated COP movement in response to medio-lateral and sagittal perturbations.** The measured medio-lateral shift of the COP in the stance foot, which characterizes the use of a lateral ankle strategy, changes from a medial to a lateral shift when the frontal plane perturbation onset varied during the gait cycle (pane a). COP shift was computed as the average difference in COP position between perturbed and unperturbed walking between 150ms-300ms after

perturbation onset. The response to the large perturbation is shown in color (blue= young, red=elderly) and to the small perturbation is shown in gray. Responses significantly different from unperturbed walking are shown in bright colors. The effect of tibialis anterior activity on COP movement in the stance foot was simulated for perturbations with onset at 37.5% of the gait cycle (pane b). A significant positive correlation was found between average reactive tibialis anterior activity and average lateral movement of the COP in the foot ($p<0.001$, R2=0.86) during the first 300 ms after perturbation (pane D). Similarly, the measured forward-backward movement of the COP in response to sagittal plane perturbations characterizes the use of an ankle strategy in the sagittal plane. A backward movement of the COP was observed in the young and older subjects response to backward perturbations (pane B) and a forward movement in response to forward perturbation (pane C). In simulation, we found that reactive tibialis anterior activity contributes the backward movement of the COP (Pane F) and soleus and gastrocnemius activity contributes to a forward movement of the COP (Pane E,G,H).

Figure 6: **Comparison between representative young and older subjects for perturbations with onset at 37.5% of the gait cycle**. Position of the Xcom with respect to the foot (pane a), reactive tibialis anterior activity (pane b) and COP motion in the foot (pane c) is shown for one representative young and older subject. The differences in reactive tibialis anterior activity and stepping strategy between the two selected subjects represent the average changes found when comparing the young and older subjects (Figure 2, 4). Reference walking data is shown in gray, response to the small perturbation in light colors (young= blue, elderly=red) and response to the large perturbation in bright colors.