Which factors drive training of standing balance in older adults?

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ABSTRACT

In many countries, aging of the population and increasing medical problems due to aging lead to increasing fall incidents. Increased medical costs and reduced life quality of a portion of the population are the side effects of this problem. Balance training is one of the most effective means to prevent falls among the older adults, but the mechanisms behind it are largely unknown, and this precludes optimization of balance training. The main accomplishment of this thesis is to improve our insight into the mechanisms underlying effective balance control and training, considering neuromuscular and biomechanical factors.

Which factors drive training of standing balance in older adults?

Leila Alizadehsaravi

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VRIJE UNIVERSITEIT

WHICH FACTORS DRIVE TRAINING OF STANDING BALANCE IN OLDER ADULTS?

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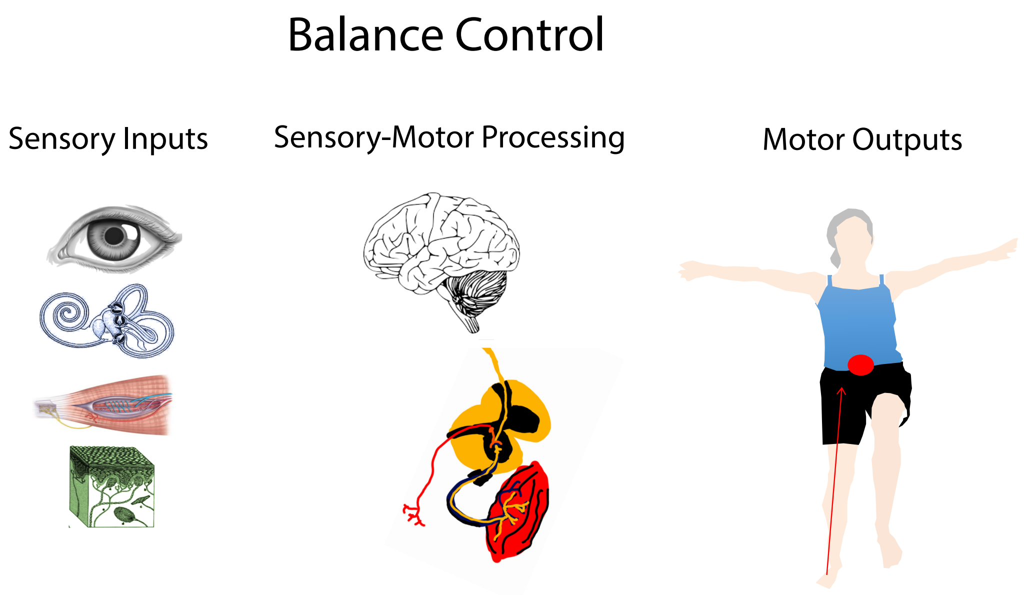
Chapter 1

The mystery of balance control in older adults



Closed-loop balance control system [1]

Balance control

Human babies learn to move from one place to another to explore their surroundings and their potential in these new places. As such, they discover new possibilities [2]. In the beginning, locomotion is difficult, and obstacles, including a limiting knowledge of one's own body (i.e., limb masses and inertias), make moving around challenging [3]. The learning process often starts with crawling, progresses via learning to stand upright with, and later without support and making supported steps, finally resulting in independent walking. The first steps challenge the baby, particularly in the single support stance phases of walking. Therefore they show shorter steps, prolonged double support, and a shorter swing time [4]. From 2 to 6 years of age, there is a rapid development of balance control [5]. Strength and coordination, along with cognitive function, develop during these years, and it takes at least seven years to perfect balance control strategies to reach levels as seen in mature adults [6]. As walking becomes easier over the years, a child learns many more dynamic motors skills, such as running, climbing, and turning. One core skill required for all of these movements is balance control.

Balance control might seem an automated process, but it requires continuous effort for humans to stabilize upright postures. Balance control is proper when one can maintain a posture and resist challenges, which might lead to a loss of balance [7]. Balance loss occurs when the body center of mass exceeds and cannot be returned within the stability limits defined by the base of support [1,8,9]. Balance functions as a closed-loop system; sensory information from visual, vestibular, and somatosensory receptors are integrated in the central nervous system, which then generates efferent motor control commands [10]. Motor commands activate the muscles, which in turn correct movements, leading to new information from the sensory systems being fed back to the central nervous system [10]. Thus, balance control requires fast integration and processing of sensory information, followed by an adequate control strategy in neural centers, controlling muscles' activations in the legs, trunk, and arms. Therefore, the effectiveness of balance control depends on several factors, such as visual, vestibular, proprioceptive acuity, muscle strength, joint mobility, and fast and adequate neural processing.

Balance control continuously evolves until we slowly start to lose it. Balance control is negatively affected by aging and several diseases [11–16]. With aging, sensory (visual, vestibular, proprioceptive, and exteroceptive sensitivity), motor (number of motor units and muscle fibers), and central nervous system (white and grey matters) functions all degrade, which leads to impaired balance control, and as a result, increases the probability of falling [1,17–20].

Thanks to adequate facilities and health care in developed countries, the older population is increasing in size, and by 2050, 40% of the EU population will be older than 55 years [17]. Falls of older adults constitute one of the leading health concerns. A sharp increase in the relative and absolute number of old and very old adults leads to an epidemic of falls with large societal costs [21]. Also, experiencing falls leads to fear of falling and avoiding physical activity, negatively impacting older adults' independence and quality of life [1]. In brief, to "Keep Control", older adults need to stay active and balanced. Understanding how balance control works, how it changes as we age, and how age-related consequences can be prevented will help older adults and their associates.

Environmental challenges

As we age, we may gradually adapt to the declining function of our organ systems. Looking at balance control as an adaptive closed-loop control system, as the system ages, it may learn to adapt to the involved organs' malfunctioning. This may be why older adults with sensory-motor degradation can still control balance in less challenging situations and reasonably known environments. However, this adaptation may not be flawless. Problems may arise when there is a change in the environment, and the control is not robust or responsive enough to deal with this. Then, a fall may occur before the system can adapt itself to fit this new condition. Some examples of such challenges could be inadequate lighting, slippery or compliant surfaces (icy surfaces, sand, or carpets), unevenness of the surface (tree roots or broken-up pavement), foot placement constraints (holes or puddles), and support surface accelerations (on busses or trains) [22,23].

Adaptations in balance control to environmental demands can be seen in sensory weighting. This is, the process by which sensory sources are integrated in a way that those sources most likely to contain the most accurate information, obtain higher importance [24]. Reweighting of information can be used to adapt to changing circumstances. For example, when walking on a compliant surface, the stance foot can be tilted while the body is oriented vertically. This implies that there is no direct association between ankle proprioception and balance control. In such conditions, vestibular and visual information are upweighted relative to proprioceptive information [25]. Older adults weigh visual input highest [11,26] and weigh proprioceptive input higher than vestibular input [27]. This may reflect differences between age groups in challenge experienced in the same task given differences in quality of the balance control system, but may also reflect more pronounced age-related degradation of the vestibular and proprioceptive systems compared to the visual system.

Sensorimotor processing for balance control was commonly thought to be executed predominantly at subcortical levels, with an important role of spinal cord circuits, based on animal studies [28]. However, even simple balance tasks require intensive cortical involvement [29–

[31]. With increasing challenges, a shift from spinal to supraspinal control has been suggested [32]. When challenged, young adults increase transmission of proprioceptive inputs to cortex [33], and their H-reflex, a marker of spinal feedback gain, is down-modulated [34]. Probably because of higher relative demands in the same task, older adults appear to rely more on corticospinal inputs for balance control than young adults [31]. For example, older adults were found to use higher prefrontal cortex activity compared to young adults even when performing a task at the same difficulty relative to age-related maximum capacity [35,36], and older adults show lower H-reflexes [34]. Older adults, however, do not seem to show adaptation of their H-reflexes with an increase in balance challenge [34], possibly reflecting that further down-modulation is not possible.

Adaptation to environmental challenges has also been observed in the tuning of muscle synergies. The central nervous system is assumed to simplify motor control and, as such, balance control, by reducing the dimensionality of its motor output in muscle synergies [37]. Such synergies can be estimated by decomposition of the activity of a number of muscles into a lower number of synergies consisting of an activation profile, reflecting the temporal pattern of activation of the synergy, and weighting factors, reflecting the extent to which the muscles are activated by the synergy [38,39]. In young adults, walking on an uneven surface or facing unpredictable perturbations widened activation profiles, which was assumed to increase robustness [40,41]. Aging appears to coincide with less consistent muscle synergies [42]. The literature on muscle synergies in older adults in relation to balance control is missing, but given the fact that the same task would be more challenging for older adults, we may expect to see wider activation profiles in older compared to young adults [43,44].

In addition, when confronting balance challenges, individuals tend to stiffen their joints by increasing co-contraction of antagonistic muscles [45]. This has been observed specifically around the ankle joints [46,47], which play a key role in balance control in standing and walking [48–51]. While both young and older adults may increase co-contraction when facing balance challenges, in the same task, older adults were found to use more co-contraction than younger adults, possibly because the task is more challenging for them [52].

From a mechanical point of view, during standing balance, two balance strategies can be used to control acceleration of the center of mass [53]. The ankle strategy involves a shift of the center of pressure induced by ankle moments. The hip or counter-rotation strategy involves redirecting the ground reaction forces through changes of the angular momentum of the body around its center of mass. This is often achieved through upper body rotation around the hip, hence in the literature, the name is called “hip strategy”. Although rotations of other segments than the upper body around the hip can also be used to change angular momentum, therefore, we will use the name counter-rotation strategy. When balance is challenged, balance strategies are adapted. With increasing perturbation magnitude and decreasing base of support, counter-rotation strategies become more dominant [54]. In addition, older adults tend to use the counter-rotation strategy more often than young adults [54,55], again possibly because for them the same task is more challenging.

Decreased balance control and increased fall risk indicate that older adults do not show optimal adaptations to environmental challenges of balance. Adaptation to environmental challenges can possibly be improved by repeated exposure to challenging situations. The above-mentioned environmental challenges can be-, and often are used as training tools to improve balance control. How fast, and transferrable skill acquisition is, and what sensory and neuromuscular mechanisms contribute to improved balance performance is still unclear.

Balance training

Balance control in individuals without diseases that affect balance can be improved in all phases of the lifespan. Balance training, specifically balance training that strongly challenges balance by means of unstable support surfaces, improves the balance performance of both young and older adults [25,56–58]. Such improvements occur faster than when training on a stable support surface [59]. Training programs often aim to prepare trainees for a sudden change in environmental demands, for instance, through training to maintain balance despite perturbations induced by waist-level pulls [60], with multidirectional platform translations [61], or platform rotations [25,62–66]. These training methods often use robot-controlled platforms, but more applicable conventional balance training equipment, such as balance boards, with several degrees of freedom and challenge levels, serve a similar purpose. Possibly, these methods are effective because they improve the ability to respond effectively to unexpected perturbations [67], slips, trips or collisions, turning, bending, and reaching.

A recent systematic review concluded that adaptations to balance training performed under strictly defined conditions are highly task-specific [68]. Nevertheless, one common aim in balance training is to increase the mobility and gait stability in older adults. Thus, finding the optimal training method to transfer balance skills to gait and daily living conditions is relevant.

While training is known to improve balance control in older adults, the optimal training duration and the mechanism underlying the balance improvement, and its transfer to tasks that challenge balance are not well-defined in older adults. Therefore, obtaining a comprehensive overview of balance control in older adults might shed light on training methods and determine training frequency and duration in the future.

Mechanisms underlying improved balance control

Balance training allows the central nervous system to regulate and retrain the body and optimize balance control. In this thesis, I aimed to understand whether training can reverse a poor balance to a good one, regardless of the organs' degradation. Alternatively, maybe there is no age-reversing process engaged in balance control after training, and it is just a matter of finding perfection in imperfection. After training, older adults could learn to modify their control, considering their impairments. They could learn to make the best use of control gains out of their impairments, to update the internal models of balance control considering their age-related degradation of sensory inputs.

Balance improvements in young adults coincide with adaptations in motor strategies, muscle activity, sensory weights, gains of neural feedback loops, cortical excitability, functional connectivity, and white and gray matter volume [25,56,57,69–71]. In line with the outcomes addressed in this thesis, it has been shown that after the balance training, young adults improved their balance performance. They showed decreased H-reflexes [57,72] and reduced the duration of co-contraction [73]. Also, older adults with poorer balance control showed higher co-contraction in the ankle joint than older adults with better balance control [74]. Furthermore, long-term training led to the use of more consistent muscle synergies while performing a balance task, which was seen in the temporal and spatial structure of the muscle activations [75]. Also, in patients with movement disorders and impaired balance, a temporal structure of the synergy showed to be broader [76] and more variable [77].

Which of these changes are determinants or consequences of improved performance in older adults? Aging might shift the control strategy from feedback to more feedforward. This way, older adults can overcome the delay in responding to unpredictable situations. However, that might need a continuous cortical and muscular effort and lead to fatigue. We aim to understand which control older adults prefer in response to varying environmental conditions and if training can alter that. Likewise, it is unclear whether foreseen improvements in balance control would transfer to daily life activities.

Main questions

This thesis addresses the following questions:

- How is balance control on surface of varying compliance different between older and young adults?
- Do older adults adapt H-reflex gains and co-contraction to varying surface compliance?

- Does balance training on unstable surfaces cause persistent improvements of balance performance and robustness in older adults?
- How fast does balance performance in older adults improve after balance training on unstable surfaces?
- Do H-reflex gains and co-contraction change after balance training on unstable surfaces in older adults?
- How are changes in balance performance related to changes in H-reflex gains, paired reflex depression, co-contraction, and synergies?
- Does balance training on unstable surfaces improve reactive balance control?
- Do the improvements in balance performance and potential mechanisms transfer to changes in gait stability?

Outline

Chapter two studies whether the reduced capacity to modulate reflex gains and co-contraction underlies balance control problems in older adults. We investigated the effect of age and varying surface compliance on spinal excitability reflected on the soleus muscle during unipedal standing. The results of this chapter led us to design a balance training program and investigate the effect of training on neural (H-reflex, co-contraction duration, paired reflex depression, and muscle synergies) and biomechanical factors underlying improvements in balance control in older adults in the next three chapters.

Obtaining a better insight into the pace of balance improvement helps to optimize balance training. In chapter three, the main goal was to reveal the changes in balance control and underlying neural and biomechanical factors after short- and long-term training. We investigated balance training effects on balance performance and robustness and potential underlying factors, including neural (H-reflex and co-contraction duration) and biomechanical factors. Also, paired reflex depression was used to give more insight into the potential peripherally induced alteration in the H-reflex mechanism after training. In chapter four, we explored the control mechanisms and changes of mechanical balance recovery strategies and muscle synergies as a result of short- and long-term training in perturbed balance.

One of the issues in balance training is transferability. This issue motivated us to study whether improvements in balance performance and robustness, as found in chapters three and four, are transferred to gait in chapter five. Thus, in this chapter, we addressed the effect of short- and long-term standing balance training on potential gait stability improvements and the neural and biomechanical factors underlying these. We investigated the changes in gait stability and muscle synergies, and kinematics during normal and narrow-base walking. The results of this chapter shed light on the transferability of improved balance after standing balance training to gait.

Finally, in chapter six, the findings of this thesis are summarized, and directions for future research and recommendations for balance training programs are given.

This thesis is unique in that it combines a cross-sectional and longitudinal approach to the study of balance control. Longitudinal studies may provide deeper insights into the determinants of proper/good balance performance and how and when training can reduce balance impairments.

Chapter 2

Modulation of soleus muscle H-reflexes and ankle muscles co-contraction with surface compliance during unipedal balancing in young and older adults

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Abstract

This study aimed to assess modulation of lower-leg muscle reflex excitability and co-contraction during unipedal balancing on compliant surfaces in young and older adults. Twenty healthy adults (ten aged 18-30 years and ten aged 65-80 years) were recruited. Soleus muscle H-reflexes were elicited by electrical stimulation of the tibial nerve while participants stood unipedally on a robot-controlled balance platform, simulating different levels of surface compliance. In addition, electromyographic data (EMG) of soleus (SOL), tibialis anterior (TA) and peroneus longus (PL) and full-body 3D kinematic data were collected. The mean absolute center of mass velocity was determined as a measure of balance performance. Soleus H-reflex data were analyzed in terms of the amplitude related to the M wave and the background EMG activity 100 ms prior to the stimulation. The relative duration of co-contraction was calculated for soleus and tibialis anterior, as well as for peroneus longus and tibialis anterior. Center of mass velocity was significantly higher in older adults compared to young adults ($p < 0.001$) and increased with increasing surface compliance in both groups ($p < 0.001$). The soleus H-reflex gain decreased with surface compliance in young adults ($p = 0.003$), while co-contraction increased ($p_{sol,ta} = 0.003$ & $p_{pl,ta} < 0.001$). Older adults did not show such modulations, but showed overall lower H-reflex gains ($p < 0.001$) and higher co-contraction than young adults ($p_{sol,ta} < 0.001$ & $p_{pl,ta} = 0.002$). These results suggest an overall shift in balance control from the spinal level to supraspinal levels in older adults, which also occurred in young adults when balancing at more compliant surfaces.

Keywords: *Balance control, postural control, spinal excitability, H-reflex, aging, co-contraction*

Introduction

In upright stance, balance is challenged by gravity and the relatively high position of the body center of mass (CoM) over a small base of support. This challenge increases with impairments in neuromuscular control resulting from age or disease [11]. But even for young, healthy individuals, maintenance of balance can become challenging when their base of support is reduced or when compliance of the surface they are standing on is increased [78,79].

In balancing on a rigid surface, moments around the ankle joint instantaneously and proportionally change the position of the center of pressure and therewith cause moments that accelerate the body center of mass [53] On a compliant surface, moments around the ankle joint change the center of pressure by moving or deforming the support surface. Consequently, the relation between the ankle moment and the center of mass acceleration is different than on a rigid surface, with changes in scaling of the effect of changes in ankle moment as well as in the temporal relation between the moment and the resulting center of mass acceleration. When standing on a compliant surface, also the relationship between sensory information from the calf muscles and the orientation of the body relative to the vertical changes. For example, with the body perfectly vertical, the ankle can still be in any orientation, as body orientation and ankle angle are decoupled. Consequently, ankle angle provides little to no information on body orientation. Balance control could potentially be adapted to such a challenge in various ways.

Considering the above, one would expect proprioceptive afference from sensors in the lower extremities to be less used when standing on a compliant surface compared to a rigid surface. In line with this, effects of calf muscle vibration, triggering muscle spindle afference, are less pronounced when standing on a compliant compared to a rigid surface [80,81]. This effect could be accounted for by sensory reweighting [25] or supraspinal suppression of motoneuron excitability. Supporting the latter mechanism, long-term training on compliant surfaces does suppress H-reflexes [57,82], but it is not clear whether immediate modulation of H-reflexes to surface compliance occurs. Experiments using a reduced base of support show indications of immediate modulations in reflex sensitivity, i.e. a negative correlations between postural demands (standing with wide or narrow base of support, prone or standing, and bipedal or unipedal stance) and H-reflex amplitudes have been reported (Koceja et al. 1995; Tokuno et al. 2009; Kawaishi and Domen 2016; Pinar et al. 2010; Kim et al. 2013). Koceja and Mynark (2000) revealed that down-modulation of the H-reflex was associated with greater postural stability, underlining the adaptive nature of this modulation. Increased postural demands also coincide with increased cortical activity [32]. These findings suggest inhibition of peripheral (spinal) control mechanisms and an increased supraspinal contribution to balance control with increasing task difficulty [31], and considering the above, this might apply specifically to increasing surface compliance. The ability to adapt balance control to surface conditions is a prerequisite to safely move through a variable environment.

Ageing causes impairments of the balance control system due to degeneration of gray and white brain matter and peripheral nerves, decreased acuity of the sensory systems and diminished muscular capacity [31,89]. Age-related reductions in H-reflex amplitudes [83] and increased cortical engagement in motor control [90], indicate an increased contribution of cortical relative to spinal inputs to balance control [31] which may reflect a bigger postural challenge in this group. Presumably, older adults need more cortical control to cope with the same task in view of age-related changes in balance control mechanisms. Older adults are also known to display increased co-contraction in postural tasks [46], which may be caused by inadequate inhibition of antagonistic muscles leading to increased joint stiffness, possibly resulting in an increased susceptibility to fall [91]. In contrast, increased co-contraction could be a compensatory strategy for impaired balance control [92], as it reduces delays in feedback control through pre-tensioning of muscle-tendon complexes [93].

In addition to experiencing an overall increase in the challenge of controlling balance, older adults appear to be less able to adapt balance control to varying environmental conditions [11]. Young adults were shown to down-modulate the soleus H-reflex between prone and standing, while older adults showed no modulation [88] or even up-modulation with postural demands [83,94].

The aim of this study was to investigate effects of varying surface compliance in mediolateral direction on single leg balance control by assessing modulation of spinal excitability and duration of co-contraction of lower-leg muscles in older compared to young adults. To the best of our knowledge, this is the first study comparing immediate adaptation in mediolateral balance control to variations in surface compliance between young and older adults. We hypothesized that balance performance decreases with increasing surface compliance and that young adults show down-modulation of spinal reflexes with increasing surface compliance. In addition, we hypothesized that older adults show less modulation of spinal reflexes and more co-contraction than young adults.

Methods

Participants

Ten young (28.2±1.3 years (Mean ± SD), 2 females, weight 70.4±16.3 Kg (Mean ± SD), height 176.2±10.0 cm (Mean± SD)) and ten older (71.4±3.9 years (Mean ± SD), 3 females, weight 79.0±11.9 Kg (Mean ± SD), height 173.3±10.0 cm (Mean ± SD)) healthy volunteers participated in this study. All younger participants were recruited through flyers distributed at Faculty of Behavioral & Movement Sciences, VU Amsterdam. All older participants were recruited through a list of older adults who previously participated in the research at our faculty, flyers, and information sharing meetings at European science night. Individuals with peripheral neuropathy, self-reported orthostatic complaints, severe visual or hearing impairments and use of medication that may negatively affect balance, were excluded. All participants provided written informed consent before participation and the procedures were approved by the ethical review board of the Faculty of Behavioral & Movement Sciences, VU Amsterdam (VCWE-2018-038).

Instruments and data recordings

Surface conditions were induced using a custom-made robot-controlled (HapticMaster, Motekforce Link Amsterdam, the Netherlands) platform with a footplate rotating in the frontal plane (Figure 2.1.a). Rotational stiffness of the footplate and damping was tunable and controlled with a simulated spring. Maximal rotation of the footplate was ±17.5°.

Full-body kinematics were acquired with one Optotrak camera array (Northern Digital, Waterloo, ON, Canada) at 50 samples/s. Six Optotrak LED marker clusters were placed on the posterior surface of the thorax, pelvis, arms and calves. The markers were tracked by the camera and anatomical landmarks were digitized in an upright posture, using a pointing probe with six markers.

Electromyographic (EMG) data were collected at 2,048 samples/s by a TMSi Refa 128-channel amplifier (TMSi, Twente, The Netherlands) data acquisition system. EMG data of the soleus, peroneus longus and tibialis anterior muscles of the stance leg were collected using bipolar, disposable adhesive surface electrodes bipolar (Ag/AgCl EMG electrodes, Ambu blue sensor N, Ambu, Ballerup, Denmark). Electrode sites were prepared by shaving the area when needed. To reduce the impedance at the skin-electrode interface, the electrode sites were cleaned with 70% isopropyl alcohol swabs. The electrode placement was chosen according to the Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) recommendations [95]. A reference electrode was placed on the lateral malleolus of the stance leg.

H-reflexes were elicited using an electrical stimulator delivering 1-ms square-wave pulses (Digitimer, DS7A UK). A large rectangular anode, roughly 6cm × 9cm, constructed of aluminum foil and conducting gel was fixed on the patella [96]. The cathode for unipolar stimulation was placed over the tibial nerve in the popliteal fossa to elicit an H-reflex in the soleus muscle. The optimal stimulation location was determined in each subject by probing the popliteal fossa with a custom-made probe for the location where the largest soleus H-reflex amplitude appeared ~25 ms after the stimulation.

Experimental procedures

Explanation and familiarization of the peripheral nerve stimulation procedure and postural conditions were provided prior to testing. To control for potential attentional and anticipatory influences on spinal reflex excitability, consistent lighting and minimal auditory input were ensured throughout the experiment. First, soleus H-reflex threshold intensity was determined using percutaneous electrical stimulation of the posterior tibial nerve during quiet, bipedal stance, and then stimulus intensity was progressively increased, with a minimum 4 sec interval, to determine the maximum H-reflex response (H_{max}) and maximal M wave (M_{max}) (Figure 2.1.b and Figure 2.1.c) [97]. During this phase, participants were instructed to visually focus on a target, while standing on both legs with their hands on their hips. Although soleus is not the most dominant muscle contributing to mediolateral balance control, it has a critical role to maintain the dynamic balancing in the frontal plane [98,99] and also soleus activation is crucial to keep the body upright while the other muscles are stabilizing the body in the frontal plane [100]. Moreover, H-reflexes can be reliably elicited in the soleus [101], therefore, we selected this muscle for studying H-reflexes.

Subsequently, ten H-reflexes were elicited using the H_{max} constant current stimulus, during unipedal stance on the balance platform at various levels of surface compliance, with three repetitions. It should be noted that during the dynamic balancing there could be changes in electrode location with respect to the nerve. Because the recruitment curve of the H-reflex is least steep around H_{max}, H-reflexes are less likely affected by such changes. Thus, by using the maximum H-reflex, we attempted to reduce errors caused by movements.

During the testing phase, participants were instructed to focus on a target in front of them, with their arms slightly abducted and their hands above the handrails of the platform, while trying to stabilize the platform in a horizontal position (Figure 2.2.a).

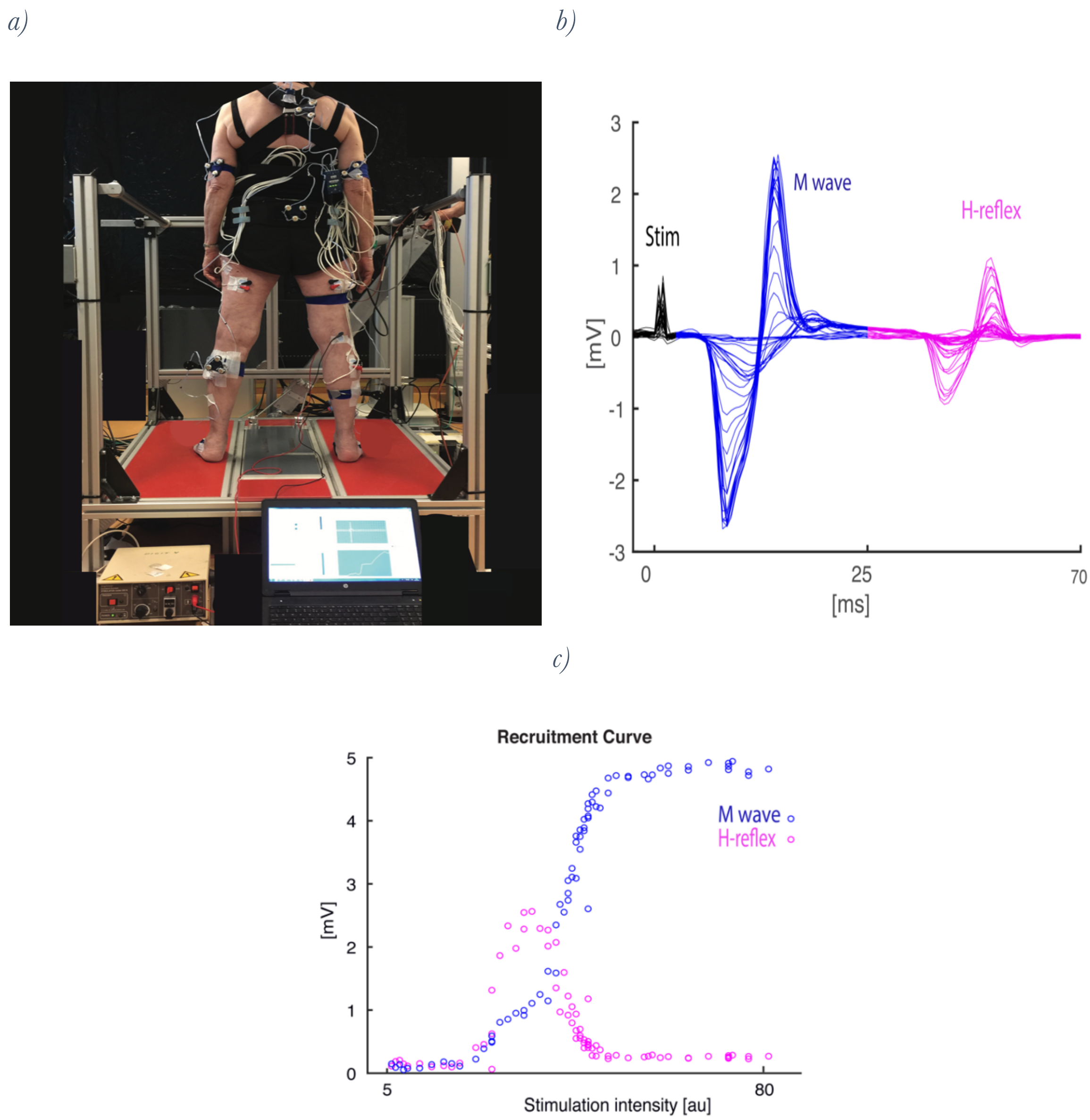

Figure 2.1: a) Experimental setup, showing a participant in bipedal stance, receiving electrical stimulation to establish the recruitment curve. b) Time series of the EMG response of the soleus muscle to the stimulation, showing traces at different stimulus intensities, each with a stimulus artefact (Stim), an M wave and an H-reflex. c) Recruitment curves, showing peak-to-peak values of M waves and H-reflexes as a function of stimulus intensity.

Participants were instructed to avoid flexing their stance leg knee during the task. A ten to fifteen seconds rest was provided between stimuli to avoid influences of post-activation depression. Thus, in total 12 balance trials were performed, of 140 seconds each, grouped into three identical blocks (randomized per subject), each consisting of four varying levels of surface compliance (rotational stiffness set at 100%, 40%, 20% and 10% of body weight multiplied by CoM height) randomized within blocks. Additionally, 4 trials of 60 seconds without stimulation at each compliance level were performed, to assess balance performance without stimulation.

Participants were given a break of 2 minutes between trials, or as long as needed to avoid any effects of fatigue.

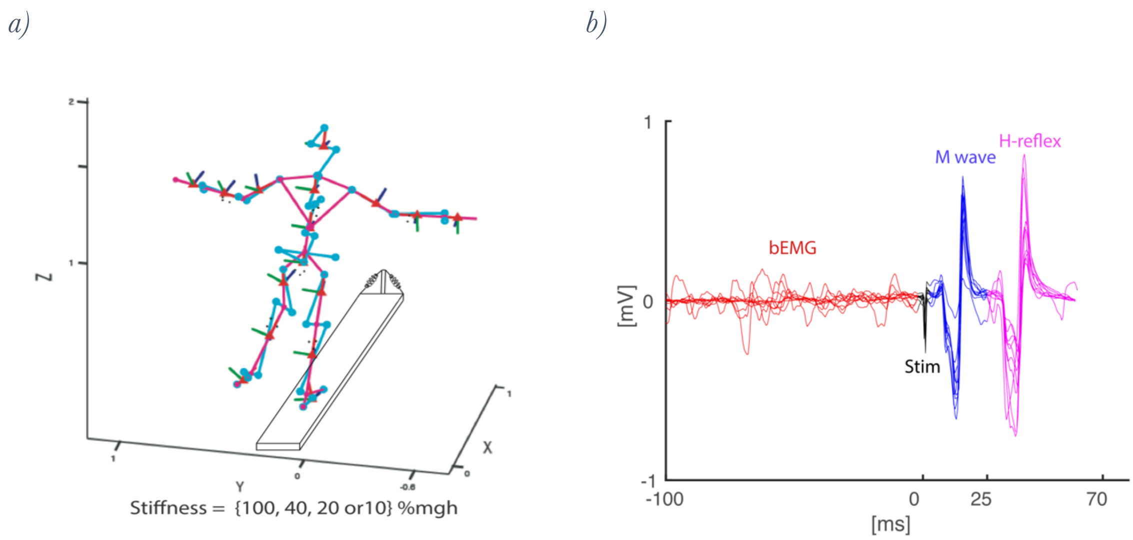

Figure 2.2: a) The kinematic model used to assess balance performance during the unipedal balance task. b) Epoched EMG data synchronized to stimulation artefacts (Stim) obtained during a balance task, showing background EMG 100 ms prior to the stimulation (bEMG), M wave and H-reflex.

Data analysis and statistics

Measures of balance performance:

Missing samples of marker coordinates were interpolated by cubic spline interpolation, and marker coordinates were low-pass filtered with a cut-off frequency of 5 Hz. The trajectories of the segments were calculated using a 3D linked segmented model (Figure 2.2.a; Kingma et al. 1996) based on the coordinates of markers and anatomical landmarks. The total body CoM position and velocity (derivative of CoM position with respect to time, vCoM) were calculated [25]. The arm segments were excluded, in view of invisibility of markers at time that participants moved their arms in front of their bodies. Supplementary material 1 chapter 2 shows that our analysis with arms included yielded similar results. The mean absolute vCoM, equivalent to the total excursion of the CoM divided by trial length, was used as a measure of balance performance (Raymakers et al. 2005; Figure 2.3). This was done both for trials during which stimulation took place, and for trials without stimulation. In trials with stimulation the results were averaged over repeated trials at an identical surface compliance.

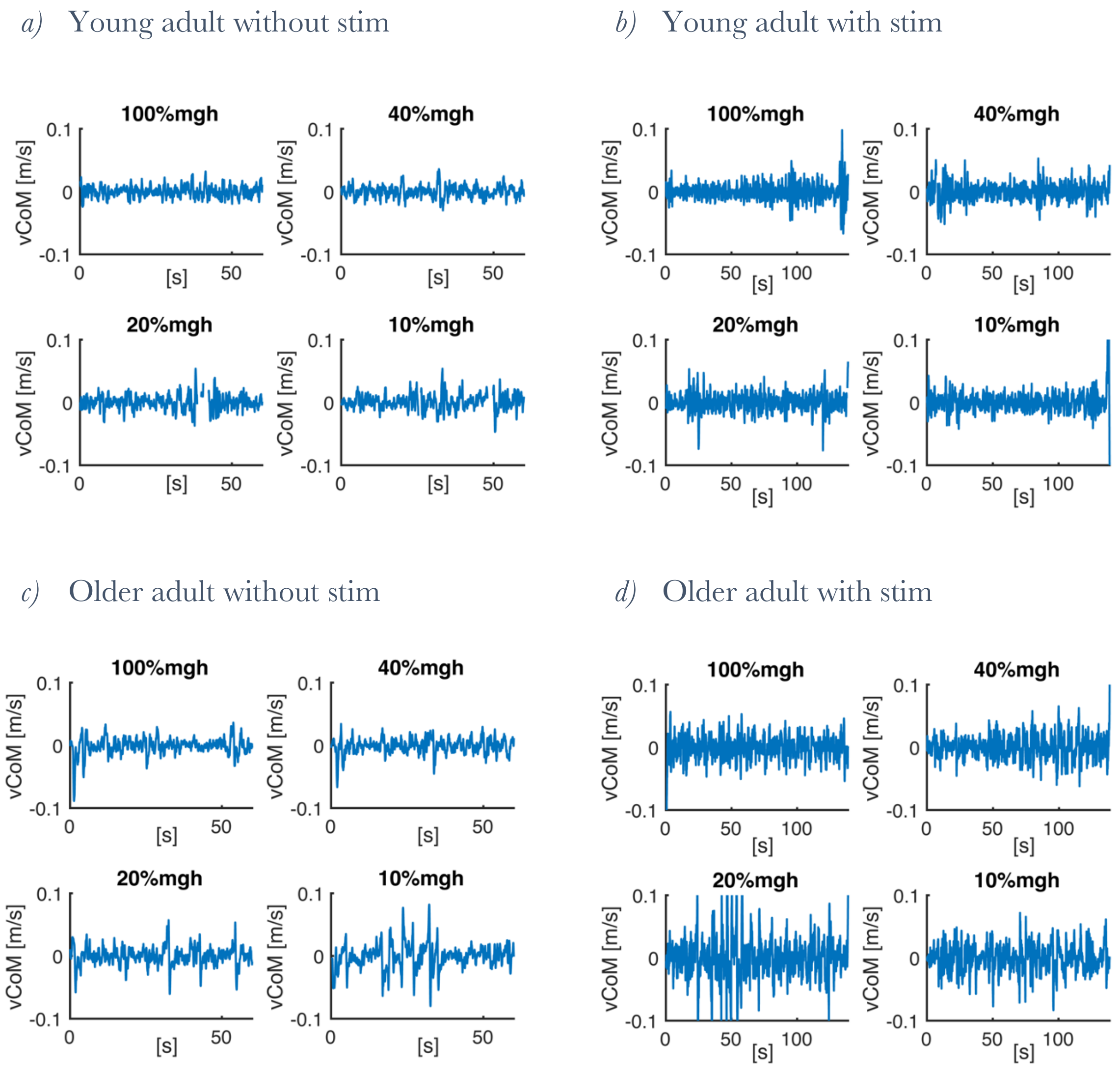

Figure 2.3: Time series of CoM velocity in one young and one older participant as a function of surface compliance in trials with and without stimulation at four levels of surface compliance (rotational stiffness set at 100%, 40%, 20% and 10% of body weight multiplied by CoM height), a) Young adult without peripheral nerve stimulation, b) Young adult with peripheral nerve stimulation, c) Older adult without peripheral nerve stimulation, d) Older adult with peripheral nerve stimulation. In both with/without peripheral nerve stimulation conditions, older adults display higher CoM velocity than younger adults, and both older and younger adults show increased CoM velocity with surface compliance.

Measures of soleus H-reflex excitability

All EMG signals were high-pass filtered at 10 Hz (2nd order bi-directional Butterworth filter) to remove movement artifacts. The amplitude of the M wave was determined as the peak to peak amplitude of the EMG from 0 to 25 ms after the stimulus artefact, the H-reflex amplitude was calculated as the peak to peak amplitude from 25 to 70 ms after the stimulus artefact. The amplitude of the background EMG (bEMG) was determined as the average rectified EMG signal over 100 ms before the stimulation (Figure 2.2.b). H/M ratio, the ratio of H-reflex amplitude and corresponding M wave amplitude, and the H-reflex gain (defined as the ratio of H-reflex amplitude divided by the bEMG [103]), were calculated. Applying bEMG normalization, we aimed to remove the effect of pre-existing motoneuron excitation [104,105]. Since the amplitude of the H-reflex linearly increases with the level of excitation of the motoneuronal pool up to 60% of maximal excitation [106,107], the H-reflex gain was considered the main outcome. Although we have not measured the maximal voluntarily activation of the soleus, excitation higher than 60% of maximal activity is not expected in the current tasks [108].

To check for consistency with previous work [86,87], we compared H-reflex amplitudes between unipedal and bipedal stance. Then we calculated the above parameters for each surface compliance condition in unipedal stance. Note that during all unipedal stance trials, the H-reflex was elicited at the stimulus intensity of H_{max} in bipedal stance.

Measure of Co-contraction

All EMG signals were first high-pass filtered at 10 Hz (2nd order bi-directional Butterworth filter) to remove movement artifacts, then rectified and low-pass filtered at 5 Hz (2nd order Butterworth). We assessed the duration of co-contraction of soleus and tibialis anterior as well as peroneus longus and tibialis anterior antagonistic muscle pairs. To this end, we determined the percentage of data points during the balance tasks without stimulation of the tibial nerve during which both muscles in a pair exceeded 10% of their maximum activation over all trials (Figure 2.4).

a) Young adult 100%mgh   *b)* Young adult 10%mgh

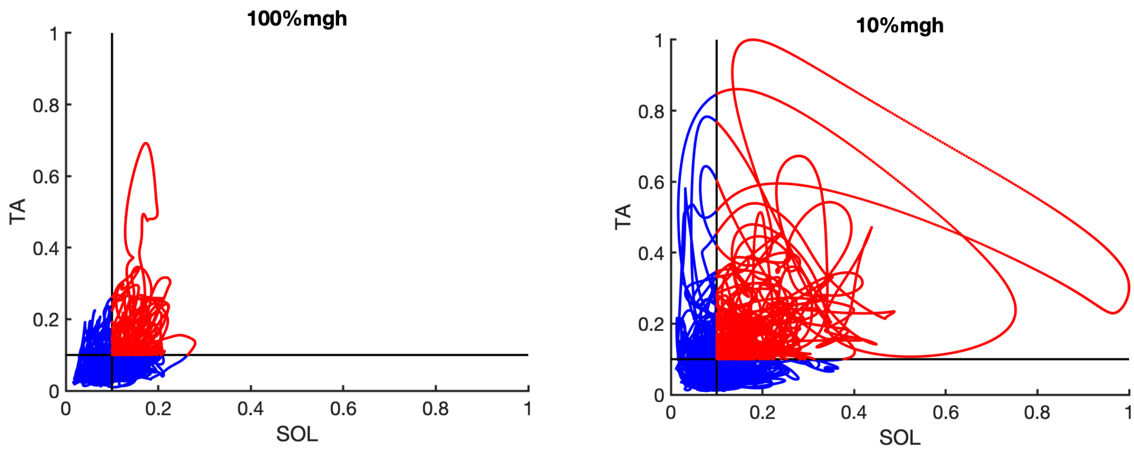

Figure 2.4: Cocontraction; results are displayed as scatter plots of tibialis anterior (TA, y-axis) and soleus (SOL, x-axis) activity of one young participant for two surface compliances, 100% and 10% of the product of body mass, gravity and the height of the CoM (mgh). All data points were normalized to the maximum activity over all trials. Data points in red indicate co-contraction (both muscles active over 10% of maximum). Data points in blue indicate no co-contraction. a) SOL TA in a young adult at 100%mgh, b) SOL TA in a young adult at 10%mgh.

Statistical analysis

All data are reported as means ± SDs. For all independent variables (absolute mean of vCoM, H-reflex excitability, co-contraction), we evaluated the effect of surface compliance and age using a 2-way mixed model ANOVA with Age (young, old) as between-subjects factor and Surface Compliance (high to low stiffness, 4 levels) as within-subjects factor. In case of interactions, post-hoc one-way ANOVAs were performed to test for effects of surface compliance within groups.

To verify that our H-reflex protocol replicated previous studies [86,87], we additionally performed a 2-way mixed model ANOVA with factors Age (young, old) and Stance Condition (bipedal to unipedal). All analyses were done in JASP version 0.9.2 (University of Amsterdam, The Netherlands), and $p<0.05$ was considered significant.

Results

Balance performance

CoM velocity in the trials without and with tibial nerve stimulation was smaller in young than older adults ($F_{(1,16)} = 12.724$, $p = 0.003$; $F_{(1,16)} = 20.013$, $p < 0.001$ respectively) and increased with increasing surface compliance ($F_{(3,48)} = 3.540$, $p = 0.021$; $F_{(3,48)} = 10.772$, $p < 0.001$ respectively) (for typical examples see Figure 2.3). No significant interaction effect of surface compliance and age group was observed ($F_{(3,48)} = 0.928$, $p = 0.435$; $F_{(3,48)} = 0.696$, $p = 0.599$ respectively). Thus, the compliant surface increased the balance challenge with decreasing stiffness, and the challenge was always greater in older than in young adults (see Figure 2.5.a and Figure 2.5.b).

a) *b)*

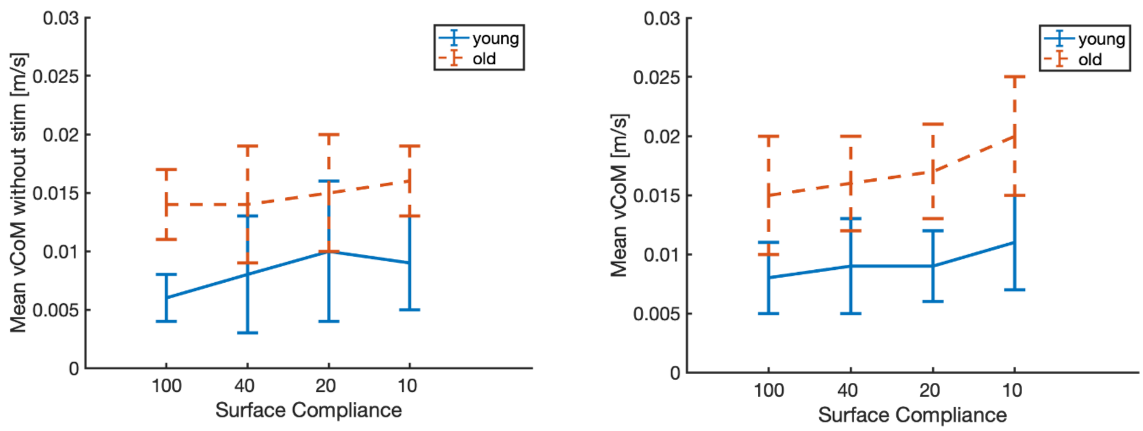

Figure 2.5: CoM velocity was higher in older than younger adults and increased with surface compliance. Displayed are group averaged values of the mean absolute CoM velocity as a function of surface compliance in trials a) without stimulation of the tibial nerve ($n_{old} = 9$, $n_{young} = 9$) and b) with stimulation of the tibial nerve ($n_{old} = 10$, $n_{young} = 8$) in young and older adults. Error bars represent standard deviations. Stiffness of the surface is expressed in % of subject weight multiplied by the height of the CoM.

Soleus H-reflex excitability

A typical example of the H-reflex responses is shown in Figure 2.2.b. The results of H-reflex amplitude, H/M ratio and H-reflex gain modulation due to surface compliance (see Figure 2.6.b, 2.6.d and 2.6.f) and stance condition (see Figure 2.6.a, 6c and 2.6.e) are presented in Tables 2.1 and 2.2 respectively.

Table 2.1: Statistical results of the comparison of H, H/M, and H-reflex gain between age groups and surface conditions.

Reflex unipedal	df1	df2	H		H/M		H-reflex gain	
			F	**p**	**F**	**p**	**F**	**p**
Surface Compliance	3	51	0.221	0.881	0.659	0.581	4.679	**0.006**
Age	1	17	10.56	**0.005**	2.926	0.105	22.42	**< .001**
Surface Compliance * Age	3	51	0.420	0.074	0.639	0.593	4.895	**0.005**

Table 2.2: Statistical results of the comparison of H, H/M, and H-reflex gain between age groups and standing conditions.

Reflex bipedal to unipedal	df1	df2	H		H/M		H-reflex gain	
			F	**p**	**F**	**p**	**F**	**p**
Stance Condition	1	18	26.45	**<0.001**	8.220	0.010	57.79	**< .001**
Age	1	18	6.435	**0.021**	0.386	0.542	12.16	**0.003**
Stance Condition * Age	1	18	1.922	0.183	0.056	0.815	6.505	**0.020**

There was no significant effect of surface compliance nor an interaction of surface compliance and age group, on H-reflex amplitude ($F(3,51) = 0.221$, $p = 0.881$; $F(3,51) = 0.420$, $p = 0.074$ respectively, see Figure 2.6.b). However, there was a significant effect of age group on H-reflex amplitude, indicating higher H-reflex amplitudes in young than older adults ($F(1,17) = 10.56$, $p = 0.005$, see Figure 2.6.b). There was no significant effect of surface compliance, age group, nor an interaction of surface compliance and age group on H/M ratio ($F(3,51) = 0.659$, $p = 0.581$; $F(1,17) = 2.926$, $p = 0.105$; $F(3,51) = 0.639$, $p = 0.593$ respectively, see Figure 2.6.d). Significant effects of surface compliance, age group and an interaction of surface compliance and age group on the H-reflex gains were found ($F(3,51) = 4.679$, $p = 0.006$; $F(1,17) = 22.42$, $p < 0.001$; $F(3,51) = 4.895$, $p = 0.005$ respectively, see Figure 2.6.f) and post-hoc testing indicated there was no significant effect of surface compliance on H-reflex gain in the older participants ($F(3,27) = 1.738$, $p = 0.186$). This is in contrast to the young adults who showed smaller H-reflex gains on more compliant surfaces ($F(3,27) =$

5.929, $p = 0.003$, see Figure 2.6.f). In summary, our hypothesis that reflex sensitivity would be down-modulated with increasing surface compliance in young but not in older adults was supported by the H-reflex gains. In addition, note that no significant M-wave variation was observed with different compliance ($F (3,51) = 1.153$, $p = 0.337$).

There were significant effects of stance condition and age group on H-reflex amplitudes, indicating smaller H-reflex amplitude in unipedal compared to bipedal stance and smaller H-reflex amplitude in older compared to young adults ($F (1,18) = 26.45$, $p < 0.001$, $F (1,18) = 6.435$, $p = 0.021$ respectively, see Figure 2.6.a). There was no significant interaction effect observed ($F (1,18) = 1.922$, $p = 0.183$). There was a significant effect of stance condition on H/M ratio indicating smaller H/M ratio in unipedal compared to bipedal stance ($F (1,18) = 8.22$, $p = 0.010$, see Figure 2.6.c), but no significant effect of age group nor an interaction of age group and stance condition on H/M ratio ($F (1,18) = 0.386$, $p = 0.542$, $F (1,18) = 0.056$, $p = 0.815$ respectively). We found smaller H-reflex gains in unipedal stance than in bipedal stance in both age groups and smaller H-reflex gains in older than young adults (($F (1,18) = 57.79$, $p < 0.001$)); $F (1,18) = 12.16$, $p = 0.003$ respectively, see Figure 2.6.e). However, a significant interaction of stance condition and age was found $F (1,18) = 6.505$, $p = 0.020$) and post-hoc tests revealed a stronger effect of stance condition in the young participants ($F (1,9) = 41.582$, $p < 0.001$) than in the older participants ($F (1,9) = 16.774$, $p = 0.003$) (Table 2.2). Overall. these results indicate reduced H-reflex sensitivity in unipedal compared to bipedal stance and decreased sensitivity in older compared to young adults, in line with previously reported findings.

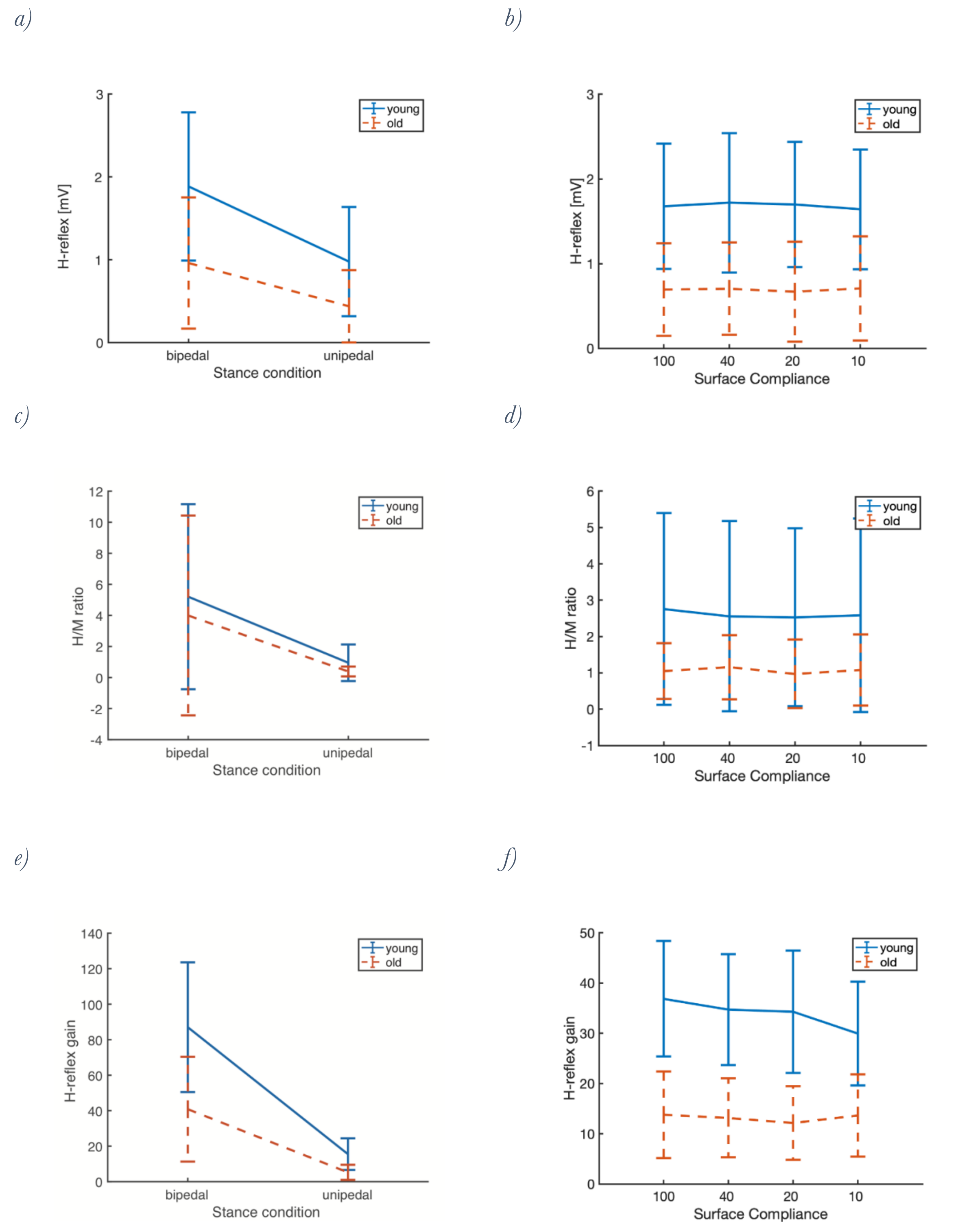

Figure 2.6: H-reflex amplitude, H/M ratio and H-reflex gain as a function of stance condition ($n_{old} = 10$, $n_{young} = 10$) in panels a, c, and e respectively and as a function of surface compliance ($n_{old} = 10$, $n_{young} = 9$) in panels b, d, and f respectively, in young and older participants. Note that decreasing stiffness from left to right on the x-axis equates increasing surface compliance. H-reflex gain was higher in younger than older adults and decreased with stance condition. H-reflex gain is down-modulated with surface compliance only in young adults.

Co-contraction

The duration of co-contraction for both muscle pairs on average was higher in older adults and increased by surface compliance, but only in young adults. The duration of co-contraction of SOL, TA and PL, TA were higher in older compared to young adults ($F_{(1,17)} = 18.37$, $p < 0.001$; $F_{(1,17)} = 14.22$, $p = 0.002$ respectively, see Figure 2.7.a and Figure 2.7.b) and increased by surface compliance ($F_{(3,51)} = 6.069$, $p = 0.001$; $F_{(3,51)} = 7.544$, $p < 0.001$ respectively, see Figure 2.7.a and Figure 2.7.b). A significant interactions of age group and surface compliance were found for the duration of co-contraction of SOL, TA and PL,TA and post-hoc testing indicated an effect of surface compliance in young participants ($F_{(3,24)} = 5.725$, $p = 0.004$; $F_{(3,24)} = 9.537$, $p < 0.001$ respectively), but not in older participants ($F_{(3,27)} = 0.909$, $p = 0.449$; $F_{(3,27)} = 0.471$, $p = 0.705$ respectively, see Figure 2.7.a and Figure 2.7.b).

a) *b)*

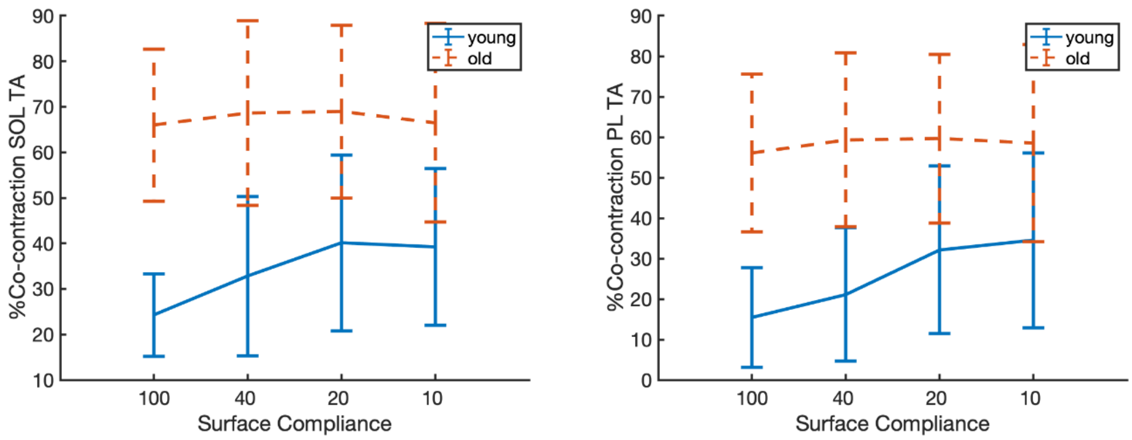

Figure 2.7: Co-contraction was not modulated with surface compliance in older adults but higher than younger adults. While in younger adults, Co-contraction increased with surface compliance. Displayed are group relative duration of co-contraction of a) soleus and tibialis anterior and, b) peroneus longus and tibialis anterior as a function of surface compliance in trials without peripheral nerve stimulation in young and older adults ($n_{old} = 10$, $n_{young} = 10$). Note that decreasing stiffness from left to right on the x-axis equates increasing surface compliance.

Discussion

We investigated differences in balance control between young and older adults on surfaces with varying compliance. In line with our hypothesis, we found that (i) balance performance decreased with increasing surface compliance in both young and older adults, (ii) older adults showed poorer balance performance than young adults, (iii) young adults showed down-modulation of H-reflex gains, although absolute H-reflex amplitudes and H/M ratios were not affected, and an increase in co-contraction with increasing surface compliance, (iv) older adults showed no modulation of H-reflex gains or co-contraction with increasing surface compliance, but lower H-reflex gains and more co-contraction than young adults in all surface conditions.

Balance performance has previously been shown to be poorer in older compared to young adults [79] and to decrease when standing on a compliant surface (foam) compared to a firm surface [79]. Similarly, our results showed a poorer balance performance, i.e. higher CoM velocities in older than in young adults and when standing on compliant surfaces in both age groups. These findings highlight that age-related impairments and surface compliance both challenge balance control and likely require adaptations in the neural control of balance to maintain stability.

One of the ways in which balance control can be altered with increasing challenge is by down-modulating spinal reflexes. A number of studies have shown down-modulation of the soleus H-reflex with increasing postural instability, such as for instance when decreasing the base of support in standing [72], or when comparing walking to standing relaxed [101] or beam walking to treadmill walking [109]. Similar down-modulation was found between bipedal and unipedal standing [86,87], as replicated in this study. Furthermore, lower H-reflexes in older compared to young adults have been found [110,111], in line with the age effects in the present study. In unipedal stance on the balance platform young adults down-modulated the H-reflex gain further with increasing challenge. As lower H-reflexes can be interpreted as a sign of reduced spinal control [92], our findings are in line with a shift in balance control from spinal to more supraspinal levels when standing on the more compliant surfaces in young adults, and more supraspinal control overall in older adults. More direct support for a shift from spinal to supraspinal control when standing on unstable surfaces was provided by Solopova et al. (2003) who showed that in adults (aged between 25-52 yrs.) TMS-evoked EMG responses of soleus muscle increased whilst, when controlled for background EMG activity, the H-reflex decreased when standing on an unstable platform compared to a stable platform. However, comparing supported versus unsupported standing, Papegaaij et al.(2016a) found decreased intracortical inhibition but no concurrent changes in H-reflexes.

Interestingly, between unipedal and bipedal stance, both age groups showed down-modulation of the H-reflex. This is in contrast with Koceja et al. (1995), who showed reduced H-reflexes in young, but not in older adults, when decreasing the base of support (prone to standing). However, these authors did find modulation of the H-reflex in a subgroup of older adults with better balance performance [83]. The older participants in the present study down-modulated their H-reflexes to some extent and, hence, may have had relatively good balance control. Why they did not further down-modulate H-reflexes in the compliant surface conditions is unknown, but it may simply be because they already had very low reflex amplitudes during unipedal stance on a fixed surface. However, an alternative explanation for the decrease in H-reflex gains across stance conditions or surface compliances could be saturation due to increased bEMG. To assess this explanation, we normalized the bEMG amplitudes to bEMG during Bipedal standing. This did not support the alternative interpretation as there were no significant age and stance effects, nor an interaction of age and stance condition, nor did we observe age or surface compliance effects, or an interaction of age and surface compliance on normalized bEMG (supplementary material 2 chapter 2).

When increasing surface compliance, young adults showed an increase in co-contraction of ankle plantar and dorsi-flexors, while older adults showed higher co-contraction overall compared to young adults. In other studies, increases in co-contraction with increasing task difficulty have been reported for young adults [93,114] as well as for older adults [115–117]. It is well known that increasing co-contraction may enhance control in some conditions [118]. However, when balancing on a compliant surface, a rigid ankle control induced by co-contraction may limit the flexibility that might be needed on such a surface. On the other hand, it may decrease response times which would benefit control [93]. Our results support an adaptive role of muscle co-contraction as we find evidence of increased co-contraction with increasing surface compliance in the young adults, as reported previously [93], but obviously this is not definitive proof of the adaptive nature of this change in control.

It is known that long-term balance training using compliant surfaces leads to improved balance in both young and older adults [119,120]. Our results suggest that such improvements would involve changes in control of the lower leg muscles and findings of decreased H-reflex gains in young adults [97] are in line with this. For older adults, it is unclear what the mechanisms behind such improved balance could be, as we found no changes in H-reflexes and co-contraction with changing surface compliance and also in long-term training no changes in H-reflex gains were found in older adults [121]. Future, long-term studies, in which H-reflexes and co-contraction along with other potential mechanisms of balance improvement are measured could elucidate the how training on compliant surfaces can improve balance control.

Limitations of the current study

This study has some limitations to be noted. First of all, the number of participants was limited. Next, In the current experimental setup, we could not use a second Optotrak camera array, to ensure uninterrupted collection of coordinates of arm markers. Consequently, we lost some kinematics data due to markers being obscured. For consistency, the arm motion data for all subjects were excluded from the analysis. However, the analysis was redone with arms included for a smaller sample size of subjects (n_{old} = 7, n_{young} = 8) without missing marker data and very similar results were obtained (as shown in the supplementary material 1 chapter 2). Another limitation of our study was that, the H-reflex is a very sensitive measure, known to be affected by several factors, such as a mental state of the participant, stimulation intensity or the muscle orientation during movement [117,122]. The recommended intensity of peripheral nerve stimulation is at 15-25% or 20-40% of M_{max} [123,124]. In line with other studies [72,125], we elicited the H-reflex at H_{max}, because the recruitment curve for the H-reflex around H-max is least steep, and thus, any potential changes in electrode location with respect to the nerve (as may occur during balancing), are likely to have less effect. Moreover, H_{max} coincided with 15-40% of M-max for most of the participants. We did not control for movement in our H-reflex analysis. A recent study used a system in which peripheral nerve stimulations were movement triggered during slackline balancing [126], which may increase reliability of outcomes. Lastly, we measured H-reflexes of the soleus, not because it has the greatest contribution in mediolateral balance control, but it does have a role in maintaining mediolateral balance [98,99] and also the H-reflex in soleus is more reliable than for other ankle muscles [101]. For a further understanding of mediolateral balance control, studying H-reflexes of other lower leg muscles may be needed.

Conclusion

In conclusion, our study reveals differences in balance control between young and older adults during a unipedal balance task and effects of surface compliance. When faced with a compliant surface, young adults decreased the soleus H-reflex gain, while increasing co-contraction. Older adults did not show such modulation in H-reflex and co-contraction.

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Chapter 3

The underlying mechanisms of improved balance after one and ten sessions of balance training in older adults

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Abstract

Training improves balance control in older adults, but the time course and neural mechanisms underlying these improvements are unclear. We studied balance robustness and performance, H-reflex gains, paired reflex depression (PRD), and co-contraction duration (CCD) in ankle muscles after one and ten training sessions in 22 older adults (+65yrs). Mediolateral balance robustness, time to balance loss in unipedal standing on a platform with decreasing rotational stiffness, improved (33%) after one session, with no further improvement after ten sessions. Balance performance, absolute mediolateral center of mass velocity, improved (18.75%) after one session in perturbed unipedal standing and after ten sessions (18.18%) in unperturbed unipedal standing. CCD of soleus/tibialis anterior increased (16%) after ten sessions. H-reflex gain and PRD excitability did not change. H-reflex gains were lower, and CCD was higher in participants with more robust balance at the last time-point, and CCD was higher in participants with better balance performance at several time-points. Changes in robustness and performance were uncorrelated with changes in CCD, H-reflex gain, or PRD. In older adults, balance robustness improved over a single session, while performance improved gradually over multiple sessions. Changes in co-contraction and excitability of ankle muscles were not exclusive causes of improved balance.

Keywords: *Balance training, center of mass velocity, co-contraction, H-reflex, paired reflex depression, motor learning, balance performance, postural balance*

Introduction

Balance control is essential to avoid falls during daily-life activities. Impaired balance control due to aging results in falls, injuries, and loss of independence in older adults [127]. To resolve this issue, it is important to understand how balance control works and when and how it improves as a result of training. Balancing requires the central nervous system to act rapidly and accurately on an array of sensory inputs [128], consisting of visual, vestibular, and tactile information, as well as proprioceptive sensory feedback [129]. Balance training leads to improved balance performance in older adults [120], observed as a reduction in mediolateral center of mass velocity during unipedal stance [79]. However, the question of how balance training induces changes in neuromuscular control remains unanswered. Hence, it is important to investigate the relation between improved balance control in older adults with changes in neural mechanisms at central and/or peripheral nervous system components.

Changes in balance control with training appear to occur at short-time scales, with substantial improvements after a single trial and over a single session [25,130,131]. Previously a rapid improvement of balance control in young adults after one session of balance training has been shown [25], while results of short-term training in older adults were inconsistent [132,133] and most studies have focused on training over several sessions spread over multiple weeks[120]. Since training effects mainly were measured before and after the entire training period only, the difference between a single session and several sessions effects of balance training in older adults is unclear.

In addition, most studies assess balance control with measures that capture balance performance, which quantifies how good people are at minimizing disturbances from an equilibrium position (often in a non-challenging condition like bipedal stance or unipedal stance on a flat surface), for instance, by measuring postural sway, in which lower values indicate better performance [63,134–136]. There are two problems with this. Firstly, subjects may choose not to minimize their sway, as higher sway values may be unproblematic and require less energy [137]. Secondly, even if subjects choose to minimize their sway, balance performance does not reflect the capability to avoid balance loss when challenged, i.e., the balance robustness, which quantifies the largest perturbation that can be resisted. Robustness has received limited attention in training literature, and if it is measured, it is mostly done in a dichotomous way (ability to perform a task, e.g., stand on one leg with eyes closed for 10 s, or not) [14]. For practical purposes, improved robustness may be more important than improved performance. While improved balance performance may not necessarily prevent falls, it may indicate improvements in balance control. Hence, we here chose to study the effects of training on both these aspects of balance control.

Age-related degenerative processes in the sensory and motor systems induce a shift from reliance on feedback control to reliance on feedforward strategies, such as co-contraction [47]. Antagonistic co-contraction can compensate for impaired sensory feedback [138]. Increasing antagonistic co-contraction when confronted with a challenging balance task is a strategy that is also used by inexperienced young adults [45,139]. Higher co-contraction in older adults with poor balance control compared to young adults with better balance control has been shown previously [74,139]. Balance training can potentially reduce levels of antagonistic co-contraction[73]. Thus, it could be expected that balance training will reduce co-contraction in older adults. We note here that many different methods have been used to assess co-contraction in the literature. In the studies mentioned above, the index of co-contraction reflected either the magnitude of antagonistic co-contraction[45] or its magnitude and duration combined [73,74,139].

Alterations of the H-reflex indicate an adjusted motoneuron output after processing of Ia afferent input at the spinal cord [140]. With age, postural modulation of H-reflexes is reduced [19,24–26,] and this may be functionally related to a declined balance performance in older adults [142]. Balance training in young adults has been reported to decrease the soleus (SOL) H-reflex [56,57,143,144]. While both young and older adults are capable of down-training the SOL H-reflex [145], it is unclear whether balance training also causes such down-regulation of the H-reflex in older adults. Unfortunately, only a few studies have addressed the effect of training on the H-reflex in older adults. Scaglioni et al. showed no changes in the H-reflex after 16 weeks of strength training in older adults [146], Ruffieux et al. found no effects of training on H-reflex after five weeks [121], and Lauber et al. showed an enhanced H-reflex after 12 weeks of alpine skiing [147]. Decreases in the H-reflex are thought to reflect a reduced effect of spinal feedback circuitry on motor control, coinciding with increased supraspinal control [56,148]. However, supraspinal mechanisms also affect the excitability of the alpha motoneuron pool and, therefore the H-reflex gain. This hampers the interpretation of the H-reflex. Therefore, measurements of paired reflex depression (PRD) were added in this study to provide an insight into peripherally induced inhibition which would more exclusively reflect changes in peripherally induced presynaptic inhibition [149–151]. The second H-reflex in PRD measurements is assumed to be influenced by the synchronous activation of the spindle's afferents during the first H-reflex. Using PRD, the influence of primary spindle afferent feedback and therefore, activation history of the Ia afferents on the motoneuron pool output can be studied[149]. Among middle-aged adults (~44 years), subjects with long-term Tai Chi practice showed better balance performance and, despite a similar H-reflex, a larger PRD[152]. These authors assumed that a reduced second H-reflex avoids overcorrection and prevents unwanted oscillations. Hence, increased PRD might be expected as a result of balance training.

The aims of the present study were twofold; first, we aimed to assess the functional benefits of one session and ten sessions of balance training in older adults. To do so, we assessed changes in balance robustness (as the duration that participants were able to keep their body balanced while surface stiffness was decreased) and balance performance (measured as the mean absolute value of the mediolateral center of mass velocity during unipedal balancing). Second, we aimed to explore the associations between the changes in balance robustness and balance performance with co-contraction duration, H-reflex gain, and PRD after one session and ten sessions of training. We hypothesized that balance robustness and performance would be improved slightly after one session, and significantly after ten sessions of training and that such improvements would be accompanied by changes, such as decreased co-contraction duration, lower H-reflex gains, and stronger PRD.

Methods

Participants

Twenty-two healthy older adults (age: 72.6 ± 4.2 years, length: 1.71 ± 0.09 m, weight: 75.6 ± 13.3 kg; mean ± SD, 11 females and 11 males) participated in this study. This is comparable to similar studies [153,154] and in accordance with a required sample size of twenty-two based on power analysis for an F test of a within factor repeated measure, assuming an effect size of 0.44 [120] and correlation among repeated measures of 0.6 ($\beta = 0.8$, G*power 3.1.9.2, Düsseldorf, Germany). To ensure participant safety and data reliability, exclusion criteria included: an inability to stand and walk for 3 minutes without walking aid, cognitive impairments (MMSE < 24), depression (GFS > 5), obesity (BMI > 30), orthopedic, neurological, and cardiovascular disease, use of medication that affects balance, and severe auditory & visual impairments. To prevent ceiling effects in balance robustness and performance and limited training gains, participants practicing sports that explicitly include balance components (e.g., Yoga, Pilates) were excluded as well [155]. To prevent obscuring any training effects, participants were asked to keep their normal activity levels in their daily life throughout the experiment. All participants provided written informed consent prior to participation, and the experimental procedures were approved by the ethical review board of the Faculty of Behaviour and Movement Sciences, Vrije Universiteit Amsterdam (VCWE-2018-171).

Experimental procedures

The protocol included an initial measurement session to determine baseline state (Pre), a measurement after one session of balance training (30 min; Post1), and after ten sessions (45 minutes per session; Post2). The protocol was concluded with a Retention assessment two weeks after the last training session. The Pre-measurements, the 30-min training session, and the Post1 measurements were performed on the same day. The measurements consisted of blocks of tests after the familiarization in the following order: assessment of balance robustness, baseline electromyography measurement (EMG, only at Pre and Post2), assessment of H-reflex, and a series of unipedal balance performance tests. During the assessments of the H-reflex and the series of unipedal balance performance tests, kinematic and EMG data were recorded. The Retention measurement consisted solely of the assessment of balance robustness (see Figure 3.1 for an overview).

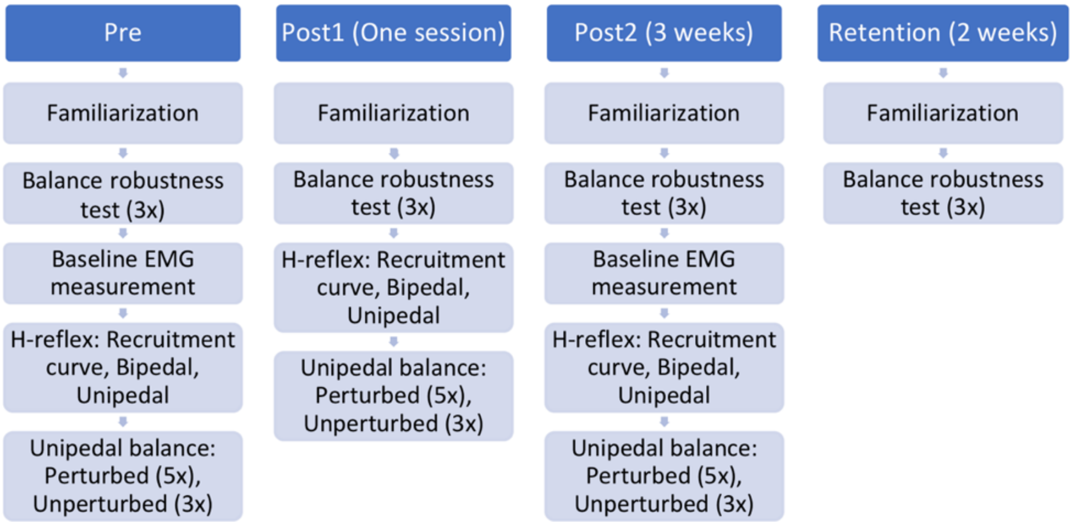

Figure 3.1: Diagram illustrating the experimental procedures.

Instrumentation and data acquisition

For all unipedal tasks, a custom-made balance platform controlled by a robot (HapticMaster, Motek, Amsterdam, the Netherlands) was used. This platform can rotate 17.5° to either direction in the frontal plane. The rotation of the platform can be controlled by the robot, simulating a tunable stiffness and damping or applying position-control. For safety reasons, the balance platform was equipped with bars in front and on both sides of the participant, and there was ample space to step off the rotating part of the platform (Figure 3.2).

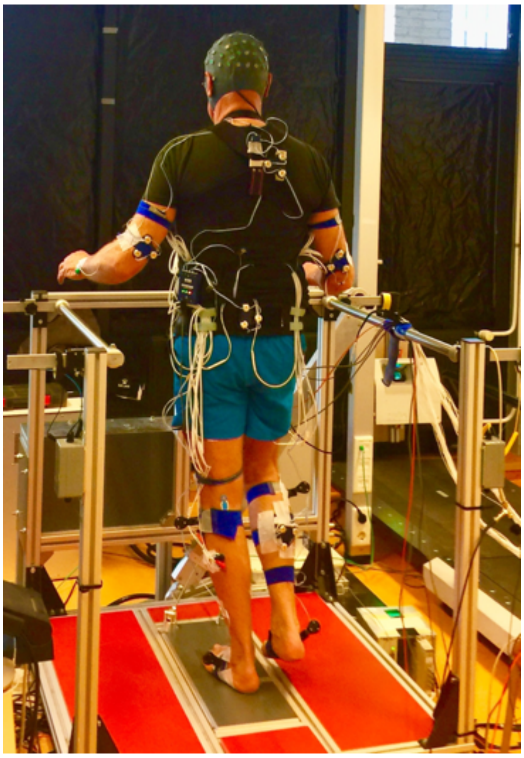

Figure 3.2: Participant in unipedal stance on the robot-controlled balance platform. This article is a part of a larger study; EEG data will be reported later.

Surface EMG data were collected from three muscles on the preferred stance leg: m. tibialis anterior (TA), m. peroneus longus (PL), and m. soleus (SOL). Bipolar electrodes were placed in accordance with the SENIAM recommendations [95]. The EMG signals were sampled at 2000 Hz and amplified using a 16-channel TMSi Porti system (TMSi, Twente, The Netherlands). The baseline EMG was measured during unipedal stance on a rigid surface. The preferred stance leg was reported by the participant prior to the experiment and confirmed by the experimenter by asking the participant to kick an imaginary soccer ball. The supporting leg was considered the preferred stance leg.

Kinematic data were obtained from 8 active marker clusters containing three markers each, placed on the posterior surface of the thorax (1), pelvis (1), arms (2), calves (2), and feet (2). The trajectories of these clusters were tracked by one Optotrak camera array (Northern Digital, Waterloo, Canada). A kinematic model of the participant was formed by relating the cluster positions to anatomical landmarks in an upright position, using a four-marker probe [156].

To elicit the H-reflex in the SOL, the tibial nerve was stimulated using an electrical stimulator (Digitimer, DS7A UK). A large diameter anode, roughly 6 × 9 cm constructed of aluminum foil and conducting gel, was fixed on the patella of the standing leg [96]. The cathode was placed over the tibial nerve in the popliteal fossa of the same leg. The optimal cathode position was determined in each subject by probing the popliteal fossa and delivering 5-10 mA stimulations to find the location that resulted in the largest SOL H-reflex amplitude ~25 ms after stimulation.

Balance robustness

Unipedal balance robustness was assessed using the balance platform. At Pre and Post2 time-points, participants were familiarized with standing on the platform on their preferred leg in two trials. In the first familiarization trial, the platform imposed ten 8° rotational perturbations at a rate of 16°/s in random direction and returned to horizontal state, every 3 s, to familiarize the subjects with perturbed unipedal balancing. For tests with varying stiffness, the rotational stiffness of the platform was normalized to percentage of *mgh* (body weight multiplied by center of mass height) of each participant, to factor out differences in participant height and mass. In the second familiarization trial, the platform was set at a stiffness of 100% *mgh* for 30 s. After familiarization and rest, the participants had to stand on their preferred leg until balance loss occurred, while the stiffness of the platform decreased stepwise every 5 s, asymptotically approximating 0 Nm/rad at the maximum trial duration of 100 s (see EQ. 1, Figure 3.3). The time an individual could stay balanced without grabbing the bar or putting down the other foot, was used to assess balance robustness. This was repeated three times, with ample rest (2-5 minutes) in between, and results were averaged over three trials.

$$\text{for time}(\mathrm{T}) = [5 * \mathrm{T} : 5 * (\mathrm{T}+1)];\ [\mathrm{s}]$$

$$\text{Stiffness}(\mathrm{T}) = \frac{100}{\sqrt{2^{\mathrm{T}}}} * \mathrm{mgh};\ \frac{\mathrm{N.m}}{\mathrm{rad}} \qquad \text{(EQ. 1)}$$

$$\mathrm{T} = 0,1,2,\ldots,\mathrm{n}\ ;\ \mathrm{n} \in \mathrm{Z}$$

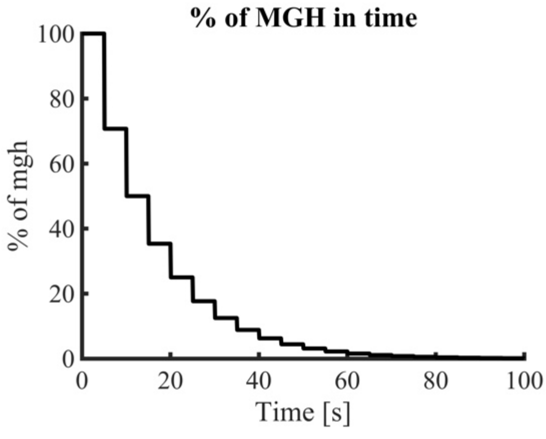

Figure 3.3: The duration of balancing in [s], and corresponding stiffness as a function of mgh (body mass times gravity times the height of the body center of mass) and time.

Unipedal balance tasks

A unipedal trial on a flat rigid surface as a baseline measurement and 2 unipedal balance tasks on the robot-controlled platform were performed: an unperturbed and a perturbed task. In the unperturbed task, the stiffness of the platform was set at a constant value. To normalize task difficulty to balance robustness, this value was set at 1.3 times the stiffness at which balance loss occurred during the assessment of balance robustness in the Pre-measurement. This task was repeated three times with two minutes rest between trials. In the perturbed task, twelve perturbations were imposed by the platform in the form of mono-phasic sinusoidal rotations either in medial or lateral direction (amplitude of 8°, angular speed of 16°/s). The perturbation direction was randomized, and the inter-perturbation duration was randomly selected between 3-5 s. This task was performed five times with two minutes rest in between trials.

H-reflexes and Paired Reflex Depression

Assessment of the H-reflex consisted of three parts: determining the recruitment curve to find H_{max} and M_{max}, measuring the H-reflex and PRD in bipedal stance, and measuring H-reflex and PRD in unipedal stance, with the intensity of the stimulator set at H_{max}. To obtain the recruitment curve, participants were subjected to low-amplitude (~5 to ~120 mA) electrical stimuli. Participants were instructed to stand still bipedally, with the feet placed at shoulder width, arms besides their body, and to focus on a target in front of them. Subsequently, 1 ms single square pulses with a minimum 4 s inter-stimulus duration were delivered to the tibial nerve at increasing amplitudes to elicit H-reflexes in the SOL and EMG data were recorded. H_{max} is the maximum peak-to-peak amplitude of the SOL EMG, between 25 and 50 ms post stimulation, and M_{max} is the maximum peak-to-peak amplitude of SOL EMG between 0 and 25 ms post stimulation.

Subsequently, H-reflex and PRD were assessed in two stance conditions [151]. In these conditions, participants were subjected to ten double-pulse stimulations of the tibial nerve. Here, inter-pulse duration was 100 ms, inter-train duration was randomized between 4-8 s, and stimulation intensity was set to the level that previously elicited the H_{max}. This stimulation protocol was delivered once in stable bipedal stance and once in unipedal stance on the balance platform, with the stiffness set at 100% mgh.

Balance training

In the first session, the participants were trained individually. The nine sessions of the 3-week training program took place in a group setting (6-8 participants). The training program was designed based on previous studies that reported improved balance and reduced fall-risk [119,157]. All training sessions were supervised by a physical therapist who ensured that the sessions remained safe, yet sufficiently challenging for all the participants. The difficulty of the exercise was manipulated by: reducing support (e.g. hand support, two-legged stance, unipedal stance), using unstable objects with varying degrees of freedom and stability, adding motor and cognitive tasks (e.g., catching a ball or passing it in changing directions), and reducing sensory information (e.g., visual fixation or eyes closed). Each session started with a short warm-up. Solely standing balance exercises, focusing on unipedal stance, were included in the training program [158] (see supplementary material 1 chapter 3). Group training sessions were 15 minutes longer than individual training sessions. Extra time was required to switch the devices between the training partners in the exercises with equipment.

Data analysis

Balance robustness

The duration the participant maintained balance, averaged over three trials, served to assess the individual's balance robustness.

Balance performance

The trajectory of the center of mass (CoM) was estimated from a full body kinematic model[102]. Balance performance was expressed as the mean absolute center of mass velocity in the mediolateral direction (vCoM).

Co-contraction duration (CCD)

Antagonistic co-contraction is the concurrent activation of antagonistic two muscles. It can be expressed as the duration, magnitude or both duration and magnitude of concurrent activation. Co-contraction was derived from three muscle pairs: SOL/TA, TA/PL and SOL/PL. EMG data were high-pass (35 Hz, bidirectional, 2nd order Butterworth) and notch filtered (50 Hz and its harmonics up to the Nyquist frequency, 1 Hz bandwidth, bidirectional, 1st order Butterworth). Subsequently, the filtered data were rectified using the Hilbert transform and low-pass filtered (40 Hz, bidirectional, 2nd order Butterworth). Finally, we determined the percentage of data points during the perturbed and unperturbed tasks at which both muscles in a pair exceeded the mean muscle activity of baseline unipedal stance. Since for Pre and Post1 time-points the measurements were performed on the same day, the same unipedal trial was used as a reference for these two time-points.

H-reflexes and Paired Reflex Depression

H-reflex gain and PRD were derived from the high-pass filtered (10 Hz, bidirectional, 2nd order Butterworth) EMG activity of the SOL. The H-reflex gain (EQ. 2) was calculated as the mean, over all pulse trains.

$$\text{Hreflex gain } = \frac{H_1}{\text{bEMG}} \qquad \text{(EQ. 2)}$$

where H1 was the maximum peak-to-peak amplitude ~25 ms after the first stimulus of the paired-pulse train and bEMG was the root-mean-square value of the EMG activity over the 100 ms prior to the pulse train. PRD was quantified as the mean relative depression of the second H-reflex relative to the first one (EQ. 3).

$$\text{PRD\%} = \frac{(H_2 - H_1)}{\text{H1}} * 100 \qquad \text{(EQ. 3)}$$

Statistics

A one-way repeated measures ANOVA was used to test the main effect of time-point (Pre, Post1, Post2, Retention) on balance robustness. Post-hoc comparisons (paired sample t-tests) were performed to investigate the effect of one session of training (Pre vs. Post1), long-term training (Pre vs. Post2), and retention (Pre vs. Retention). In addition, Post1-Post2 and Post2-Retention were compared to obtain insight into the changes over the short- and long-term and in retention.

Two-way repeated-measures ANOVAs were used to identify main effects of time-point (Pre, Post1, Post2) and condition (perturbed/unperturbed or bipedal/unipedal) on vCoM, CCD, H-reflex gain, and PRD. When the assumption of sphericity was violated, the Greenhouse-Geisser method was used. Post-hoc analyses (paired samples t-test) were performed to investigate the effect of one session (Pre vs. Post1) and ten sessions (Pre vs. Post2) of training when a main effect of Time-point or an interaction of Time-point x Condition was observed. For all post-hoc analyses, Holms' correction for multiple comparisons was applied.

Balance performance and the response to training are heterogeneous in older adults[120]. Therefore, cross-sectional and longitudinal correlation analyses were performed to gain more insight into which (changes in) co-contraction, H-reflexes, and PRD were related to (changes in) balance robustness and balance performance. As cross-sectional analyses, the correlations between balance robustness (duration) and CCD (averaged over perturbed and unperturbed trials) for all muscle pairs, H-reflex, and PRD were calculated. Moreover, the correlations between balance performance (vCoM) and the CCD for all muscle pairs in perturbed and unperturbed trials and between balance performance (vCoM) and H-reflex gains and PRD during unipedal and bipedal stance were calculated for the three time-points. For longitudinal analyses, the correlations between changes in the same parameters after one session and ten sessions of training were calculated. In view of outliers, Spearman's correlation (r) coefficients were calculated. In all statistical analyses, α=0.05 was used.

Only in balance robustness, all participants were included in the analysis. For all other analyses twenty-one participants were included because one participant was not able to fully perform the balance performance trials.

Results

Balance robustness

Balance robustness (duration of balancing) increased as a result of balance training ($F_{1.955,41.060}$= 10.637, $p < 0.001$). The mean duration of balancing increased after one session of training ($t = 3.325$, $p = 0.006$, Figure 3.4). While, the duration remained unchanged between Post1-Post2 and Post2-Retention ($t = -1.257$, $p = 0.427$; $t = -0.57$, $p = 0.571$, respectively; Figure 3.4), ten sessions of training and retention showed higher robustness than Pre time-point ($t = -4.582$, $p < 0.001$; $t = -5.151$, $p < 0.001$, respectively; Figure 3.4). Overall, these results indicate a rapid improvement in balance robustness after only one session of training, with no further improvement after the subsequent nine training sessions.

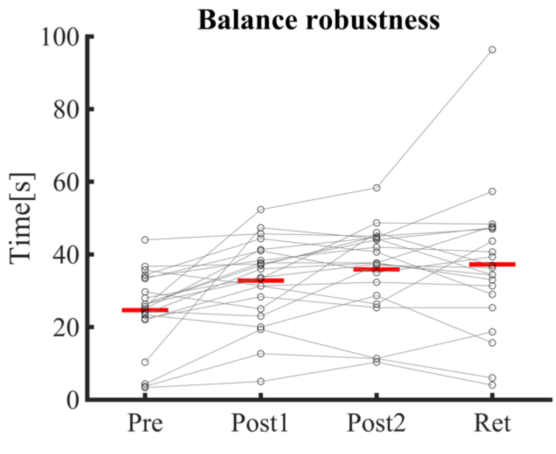

Figure 3.4: Balance robustness at different time-points, expressed as the duration of maintaining balance under gradually decreasing surface stiffness.

Balance performance

Perturbed and unperturbed

Balance training led to an increase in balance performance (i.e. decreased vCoM, Figure 3.5, Time-point effect, $F_{1.533,30.655}$= 10.598, $p < 0.001$). Participants showed larger vCoM in perturbed compared to unperturbed standing (Figure 3.5.a & 3.5.b, Condition effect, $F_{1,20}$ = 58.285, $p < 0.001$). Additionally, there was a significant interaction of time-point and condition on vCoM ($F_{2,40}$= 5.242, p = 0.01). Post-hoc analysis showed that one session of training decreased vCoM in the perturbed condition, but did not change vCoM in the unperturbed condition (t = 3.35, p = 0.011 and t = 1.193, p = 0.715, respectively). On the other hand, ten sessions of training, changed vCoM significantly in both perturbed and unperturbed condition (t = 5.206, $p < 0.001$; t = 3.394, p = 0.011, respectively; Figure 3.5.a & 3.5.b), even though there were no significant changes in vCoM between Post1 and Post2 measurements in perturbed and unperturbed conditions (t = 1.439, p = 0.783; t = 1.718, p = 0.553, respectively; Figure 3.5.a & 3.5.b).

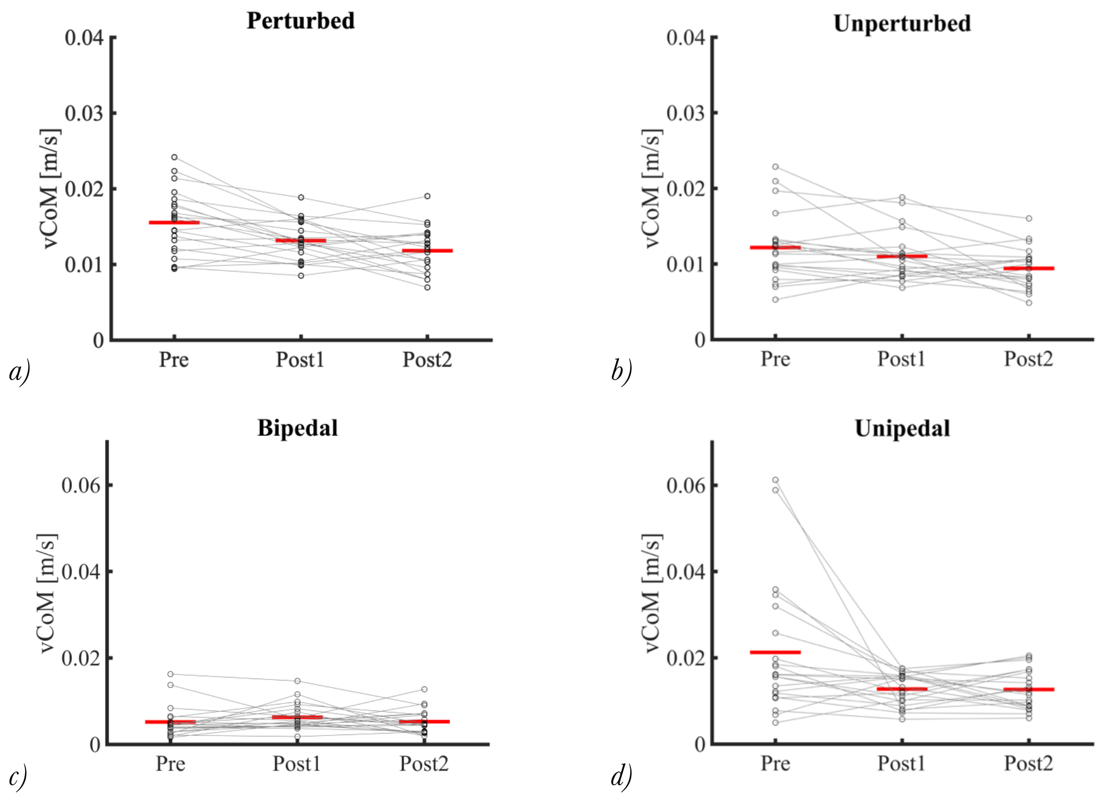

Figure 3.5: The mean absolute center of mass velocity in mediolateral direction at all three measured time-points. a) in the perturbed condition, b) in the unperturbed condition, c) in H-reflex bipedal stance condition, and d) in H-reflex unipedal stance condition. Circles and connecting lines represent individual results. The red lines indicate averages across subjects.

Bipedal and unipedal (H-reflex trials)

In one participant at time-point Pre, during the bipedal H-reflex measurement, a marker on the left arm was not visible. Therefore, for this participant the arms were excluded in calculating CoM trajectories. There was a significant effect of Time-point on balance performance ($F_{1.163,23.267}$= 5.233, p = 0.027). Participants showed smaller vCoM in bipedal compared to unipedal standing (Figure 3.5.c & 3.5.d, Condition effect, ($F_{1,20}$= 63.924, p < 0.001). There was a significant interaction of Time-point x Condition on vCoM ($F_{1.249,24.974}$= 6.237, p = 0.014). Post-hoc analysis showed that vCoM decreased only in the unipedal condition, after the one session and ten sessions of training (t = 4.101, p = 0.001; t = 4.147, p = 0.001, respectively), even though there were no significant changes between Post1-Post2 time-points (t = 0.046, p = 1).

Duration of Co-contraction

The CCD of the SOL/TA muscle pair was affected by time-point (Table 3.1). Post-hoc comparison showed that the CCD was not changed after the one session but had increased after ten sessions of training (t = 1.623, p = 0.112; t = - 2.372, p = 0.045, respectively; Figure 3.6.a). No effects of Time-point and Condition, nor an interaction were observed for the other muscle pairs (Table 3.1). Overall our results showed no changes in SOL/TA CCD after one session of training but an increased SOL/TA CCD after ten sessions of training.

Table 3.1: Results of repeated-measures ANOVA of the duration of co-contraction of three muscle pairs, in perturbed and unperturbed standing at three Time-points of Pre, Post1, and Post2. Bold numbers indicate a significant effect.

Paradigm	Muscles	Time-point			Condition			Time-point*Condition		
		df	**F**	**p**	**df**	**F**	**p**	**df**	**F**	**p**
CCD	**SOL TA**	**1.512, 30.242**	**8.073**	**0.003**	1,20	0.416	0.526	2,40	0.298	0.726
	TA PL	1.148, 22.952	1.285	0.275	1,20	0.005	0.944	1.567, 31.336	1.018	0.370
	SOL PL	1.088,37.665	0.522	0.492	1,20	0.616	0.441	2,40	0.762	0.473

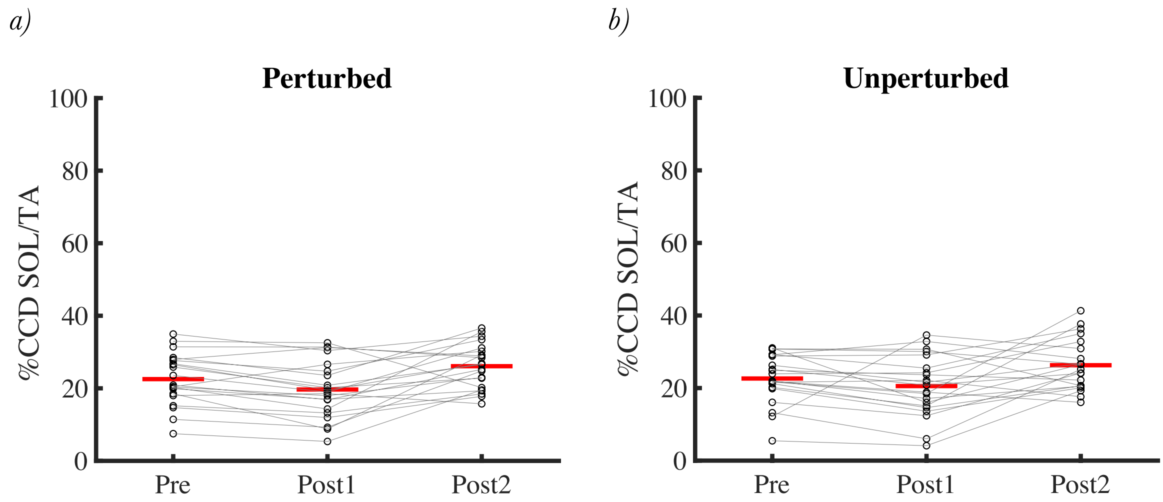

Figure 3.6: Co-contraction index at three time-points in a) perturbed and b) unperturbed standing, for the muscle pairs SOL/TA. Circles and connecting lines represent individual results. The red lines indicate averages across subjects.

Reflexes

There was no effect of Time-point, nor an interaction effect of Time-point x Condition on H-reflex gains ($F_{1.567,31.344}$= 0.467, p = 0.585, and $F_{2,40}$ = 1.859, p = 0.169, respectively; Figure 3.7). H-reflex gains were significantly higher in bipedal compared to unipedal stance ($F_{1,20}$= 26.549, p < 0.001). Similarly, there was no effect of Time-point, nor an interaction effect of Time-point x Condition, on PRD ($F_{2,40}$= 1.043, p = 0.360, and $F_{2,40}$= 0.204, p = 0.802, respectively; Figure 3.8) but PRD was stronger in bipedal compared to unipedal stance ($F_{1,20}$= 39.613, p < 0.001). Overall our results did not show any changes in the reflexes as a result of training.

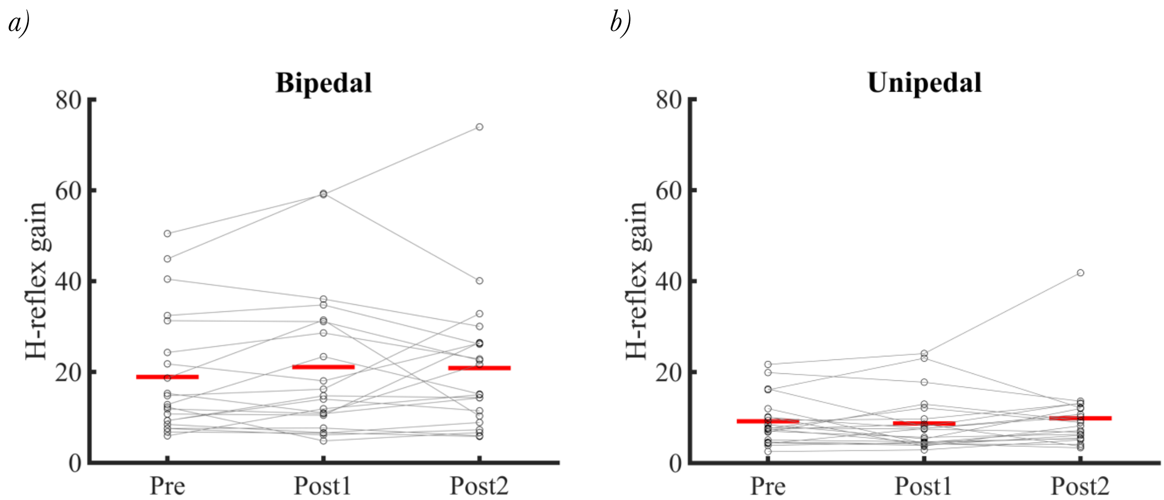

Figure 3.7: H-reflex gains at three time-points a) shows the reflex gain for the bipedal condition b) shows the reflex gain for the unipedal condition. Circles and connecting lines represent individual results. The red lines indicate averages across subjects.

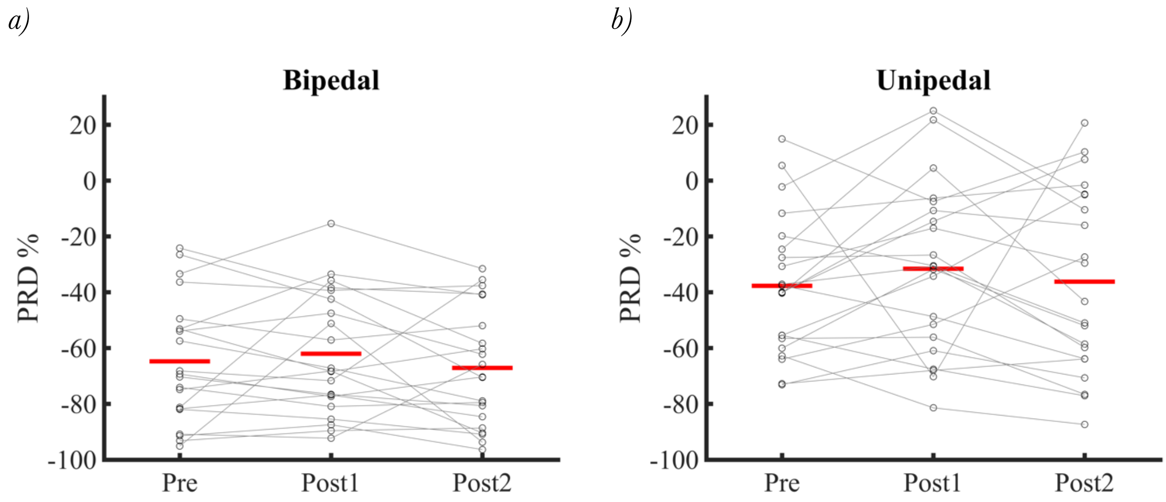

Figure 3.8: Paired reflex depression at three time-points. The paired reflex depression is displayed for a) the bipedal condition, and b) unipedal condition. Circles and connecting lines represent individual results. The red lines indicate averages across subjects.

Associations of balance robustness with co-contraction and reflexes

All correlation results are shown in Tables 3.2 and 3.3. For co-contraction, the average of the perturbed and unperturbed SOL/TA CCD was positively correlated with balance robustness at time-point Post2 ($r = 0.564$, $p = 0.007$). No correlations were observed between changes after one session or ten sessions of training. For reflexes, H-reflex gains in unipedal stance were negatively correlated with balance robustness (duration) at time-point Post2 ($r = -0.585$, $p = 0.005$). No correlations were observed between changes after one session or ten sessions of training.

Table 3.2: Results of the correlational analysis between co-contraction (averaged over perturbed and unperturbed trials), reflexes in bipedal and unipedal with balance robustness (duration) at each Time-point. Bold numbers indicate a significant effect.

	Pre Balance robustness		Post1 Balance robustness		Post2 Balance robustness	
	r	p	r	p	r	p
CCD TAPL	-0.183	0.425	0.326	0.148	0.037	0.873
CCD SOLTA	-0.015	0.948	0.115	0.617	**0.564**	**0.007**
CCD SOLPL	-0.277	0.221	-0.114	0.621	-0.229	0.317
H-reflex gain Bi	0.193	0.398	-0.183	0.426	0.050	0.827
H-reflex gain Uni	0.183	0.425	-0.044	0.849	**-0.585**	**0.005**
PRD Bi	-0.363	0.105	0.119	0.605	0.114	0.621
PRD Uni	-0.384	0.086	-0.063	0.783	0.113	0.625

Table 3.3: Results of the correlational analysis between the changes of co-contraction (averaged over perturbed and unperturbed trials), changes of reflexes in bipedal and unipedal with changes of balance robustness (duration) after one session and ten sessions of training.

	One session Δbalance robustness		Ten sessions Δbalance robustness	
	r	**p**	**r**	**p**
ΔCCD TAPL	0.076	0.743	-0.380	0.089
ΔCCD SOLTA	0.302	0.182	-0.013	0.957
ΔCCD SOLPL	0.085	0.713	-0.392	0.079
ΔH-reflex gain Bi	-0.162	0.481	0.015	0.948
ΔH-reflex gain Uni	0.296	0.191	0.102	0.657
ΔPRD Bi	-0.352	0.116	0.053	0.819
ΔPRD Uni	-0.005	0.982	0.100	0.665

Associations of balance performance with co-contraction and reflexes

All correlation results are shown in Tables 3.4 and 3.5. For co-contraction duration, at time-point Pre, and Post2, SOL/TA CCD was negatively correlated with vCoM in perturbed standing ($r = -0.441$, $p = 0.046$; $r = -0.471$, $p = 0.032$, respectively), and at time-point Pre, TA/PL CCD was negatively correlated with vCoM in unperturbed standing ($r = -0.453$, $p = 0.040$). Negative correlations indicate that higher duration of co-contraction was associated with better performance (lower sway velocity). No correlations were observed between changes after one session or ten sessions of training (Table 3.5).

For reflexes, at time-point Post1, PRD was positively correlated with vCoM in bipedal stance ($r = 0.583$, $p = 0.006$), indicating that stronger PRD was associated with better performance. No correlations were observed between changes after one session or ten sessions of training (Table 3.5).

Table 3.4: Results of the correlational analysis between co-contraction with vCoM in perturbed and unperturbed, and between reflexes in bipedal and unipedal with vCoM in bipedal and unipedal stance at each Time-points of Pre, Post1, and Post2. Bold numbers indicate a significant effect.

	Perturbed					
	Pre vCoM		**Post1 vCoM**		**Post2 vCoM**	
	r	**p**	**r**	**p**	**r**	**p**
CCD TAPL	-0.306	0.176	-0.426	0.055	0.040	0.863
CCD SOLTA	**-0.441**	**0.046**	-0.340	0.131	**-0.471**	**0.032**
CCD SOLPL	0.270	0.235	-0.071	0.758	0.154	0.501
	Unperturbed					
CCD TAPL	**-0.453**	**0.040**	-0.274	0.228	0.168	0.462
CCD SOLTA	-0.268	0.237	-0.285	0.208	-0.189	0.408
CCD SOLPL	0.277	0.221	0.194	0.395	0.288	0.204
	Bipedal					
H-reflex gain Bi	0.066	0.775	-0.137	0.550	0.310	0.170
PRD Bi	-0.375	0.094	**0.583**	**0.006**	-0.275	0.226
	Unipedal					
H-reflex gain Uni	0.284	0.210	-0.009	0.970	0.185	0.418
PRD Uni	-0.045	0.845	0.305	0.178	-0.088	0.702

Table 3.5: Results of the correlational analysis between changes of co-contraction with changes of vCoM in perturbed and unperturbed, and changes of reflexes in bipedal and unipedal with changes of vCoM in bipedal and unipedal stance after one session and ten sessions of training

	One session ΔvCoM		Ten sessions ΔvCoM	
	r	**p**	**r**	**p**
	Perturbed			
ΔCCD TAPL	-0.089	0.698	-0.258	0.256
ΔCCD SOLTA	-0.003	0.988	0.001	0.997
ΔCCD SOLPL	0.044	0.85	0.039	0.86
	Unperturbed			
ΔCCD TAPL	-0.002	0.993	0.022	0.925
ΔCCD SOLTA	0.306	0.176	0.085	0.711
ΔCCD SOLPL	0.374	0.095	0.203	0.373
	Bipedal			
ΔH-reflex gain Bi	-0.367	0.101	0.347	0.124
ΔPRD Bi	-0.115	0.616	-0.386	0.085
	Unipedal			
ΔH-reflex gain Uni	-0.414	0.063	0.234	0.306
ΔPRD Uni	0.144	0.531	-0.039	0.868

Discussion

We investigated the functional benefits and neural mechanisms associated with functional benefits of one session and ten sessions of balance training in older adults. We found that only one session of balance training increased older adults' balance robustness. Extra training sessions did not further improve but maintained the acquired robustness. In addition, balance performance in perturbed unipedal balancing was improved after only one training session, again with no further improvement over subsequent training sessions. Performance in unperturbed unipedal balancing, significantly improved over a ten sessions training period, in line with previous studies [120].

In terms of challenge, the perturbed balance performance test and also the unipedal test during H-reflex stimulation can be considered intermediate to the unperturbed balance test and the test for robustness. We suggest that robustness and perturbed performance outcomes are mainly limited by the ability to deal with near balance loss, while the unperturbed balance test reflects the ability to minimize sway in a situation where balance loss is not likely to occur. The fast changes in the ability to recover balance would be in line with results on perturbation training [130,131]. Overall, this suggests that balance training can increase robustness rapidly, while ten sessions of training refines balance performance and maintains the acquired balance robustness and performance. Given the functional relevance of balance robustness, this finding would put into question the predominant use of balance performance in conditions with a low challenge as outcome measures of training. We note here that balance performance during bipedal standing was not affected by training.

Contrary to our hypothesis, co-contraction was not decreased after balance training; one session of training did not change the co-contraction duration, and ten sessions of training even led to an increased co-contraction duration of SOL/TA. Moreover, cross-sectional correlation analysis showed higher co-contraction duration was correlated with higher balance robustness and performance. Co-contraction may be an adaptation, and training could reduce the need for it - older adults show more co-contraction than young adults [74]. But, training also could increase the use of this adaptation. Co-contraction of antagonistic muscles has been shown to increase joint stiffness and serve a zero-delay corrective response to unexpected disturbances in challenging motor tasks [159]. In addition, co-contraction may reduce electromechanical delays by pre-tensioning tendons, and as such improved feedback control [93] and co-contraction may improve feedback response by allowing dual control of agonist and antagonistic muscles [160]. Therefore, older adults may increase co-contraction to enhance balance control. However, longitudinal analysis did not show any correlation between the changes in co-contraction duration and changes in balance robustness or performance. Therefore, it seems that increased co-contraction duration is not the mechanism underlying improved balance after one session or ten sessions of training. Possibly, training causes some individuals to use co-contraction more, whereas it reduces the need for co-contractions in others.

Also, in contrast with our hypothesis, neither one session, nor ten sessions of training affected H-reflex gains or PRD. In line with previous studies [86,87,139], H-reflex gains decreased when going from bipedal to unipedal stance. This has been suggested to help in dealing with the higher postural demand of unipedal stance [161], where monosynaptic stretch reflexes may fail to contribute to maintenance of balance. However, we found stronger PRD in bipedal than unipedal stance. It has been suggested that the inhibitory effect of the first H-reflex stimulus is less when more background afferent discharge is present, which could explain the difference between unipedal and bipedal stance [149]. Alternatively, PRD may be affected by descending pathways projecting onto spinal interneurons, resulting in a larger second H-reflex (less depression) in unipedal compared to bipedal stance [162]. Functionally this decreased depression could act to facilitate responses to external perturbation, but this would be at odds with the decreased gain of the first H-reflex. Cross-sectional analyses showed that, in unipedal stance, smaller H-reflex gains were correlated with higher balance robustness, and stronger PRD in bipedal stance correlated with better balance performance. Longitudinal correlational analyses did not show any significant correlation between the neuromuscular mechanisms and the performance or the robustness. All in all, these data support that lower excitability in response to type 1a afference and stronger suppression of responses to such input is beneficial for balance control, in line with outcomes of studies in middle-aged adults [152], but changes in H-reflex sensitivity or depression do not appear to account for the effect of training.

Limitations

Since multiple randomized controlled trials have shown the efficacy of balance training in older adults, the present study was done without a control group [120]. This implies however, that we cannot exclude that some of our findings were due to repeated testing, which in itself could be seen as a form of training. The finding that balance robustness did not drop two weeks after the last training session and hence five weeks after initial testing indicates that the improvement was a result of learning. Second, our hypothesis that co-contraction duration will decrease after the balance training in older adults, was based on the findings from our previous study, where we found higher co-contraction duration in older adults compared to younger adults. Hence, we used the method presented in the current study comparable with our first study, which takes the duration of co-contraction as a percentage of when muscles are active, as determined from a reference activation. This EMG baseline measurements itself could be influenced by training, and higher co-contraction duration after the ten sessions of training resulted in our study could be simply due to lower baseline measurement at Post2 Time-point. Also, co-contraction could be an adaptation mechanism and training could reduce or increase it, considering the impairment and the task. Third, for reflex measurements it is generally recommended to elicit H-reflex between 15-40% of M_{max} [123,124], while we elicited H-reflex at H_{max}, in line with our previous study. However, for 20 out of 22 participants H_{max} was less than 40% of M_{max} (see supplementary material 2 chapter 3). Lastly, we calculated a large amount of correlations, and did not apply a correction for multiple testing while doing so. Hence, our results should be considered as explorative, and future, confirmative studies should be undertaken to confirm our findings.

Perspective

Previous studies showed improved balance performance as a result of balance training in both young and older adults [25,120,163]. In young adults improved balance control has been shown to be accompanied with decreased H-reflexes [57] and decreased co-contraction [73]. In older adults, the mechanisms underlying improvements in balance performance and robustness after the training remain unclear. Our results indicate that one session of training improves balance robustness, while ten sessions of training led to a better balance performance with no further improvement in but potentially contributing to retention of balance robustness. While co-contraction duration was correlated to balance performance cross-sectionally, the neural mechanisms underlying balance improvement after one or ten training sessions were not exclusively the ones we studied here (i.e., co-contraction duration, H-reflex gain and peripherally induced inhibition measured with PRD).

Acknowledgments

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Chapter 4

Balance training improves feedback control of perturbed balance in older adults

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Abstract

Recovering balance after perturbations becomes challenging with aging, but an effective balance training could reduce such challenges. In this study, we examined the effect of balance training on feedback control after unpredictable perturbations by investigating balance performance, recovery strategy, and muscle synergies. We assessed the effect of balance training on unipedal perturbed balance in twenty older adults (>65 years) after short-term (one session) and long-term (3-weeks) training. Participants were exposed to random medial and lateral perturbations consisting of 8-degree rotations of a robot-controlled balance platform. We measured full-body 3D kinematics and activation of 9 muscles (8 stance leg muscles, one trunk muscle) during 2.5 s after the onset of perturbation. The perturbation was divided into 3 phases: phase1 from the onset to maximum rotation of the platform, phase 2 from the maximum angle to the 0-degree angle and phase 3 after platform movement. Balance performance improved after long-term training as evidenced by decreased amplitudes of center of mass acceleration and rate of change of body angular momentum. The rate of change of angular momentum did not directly contribute to return of the center of mass within the base of support, but it reoriented the body to an aligned and vertical position. The improved performance coincided with altered activation of synergies depending on the direction and phase of the perturbation. We concluded that balance training improves control of perturbed balance, and reorganizes feedback responses, by changing temporal patterns of muscle activation. These effects were more pronounced after long-term than short-term training.

Keyword: *balance training, balance control, feed forward, feedback, counter-rotation, recovery, synergy, aging*

Introduction

In theory, the nervous system can use two control mechanisms to recover balance after a perturbation [164]. Reactive or feedback control, occurs after a perturbation and is the only mechanism available when the nervous system has no prior knowledge of a perturbation [165,166]. Anticipatory or feedforward control is based on expectations of a perturbation, and aims to minimize the impact of the perturbation on balance by changing joint orientation or stiffness prior to a perturbation [167]. Depending on a perturbation's direction and magnitude, feedforward control is not always sufficient for balance control, and then feedback control comes into play to regain balance. Effective feedforward control minimizes the effect of perturbations and reduces the need for feedback control [168,169].

Three movement strategies are well known to contribute to feedback control of balance after perturbations: the ankle, counter-rotation, and stepping strategies [170]. The stepping strategy aims to displace or expand the base of support beyond the projection of the center of mass by stepping or grabbing a handhold. It is usually seen as a last resort reflecting poorer balance control, and older adults use it more than younger adults [171–173]. The ankle strategy aims to accelerate the center of mass towards the base of support through a shift of the center of pressure, the point of application of the ground reaction force, generated by ankle moments [174]. The counter-rotation strategy aims to accelerate the center of mass towards the base of support through horizontal ground reaction forces generated by changes in the angular momentum of body segments relative to the center of mass [25,170,174]. Thus, these strategies can be differentiated by distinct kinematics and kinetics but also by distinct patterns of muscle activation reflected in distinct muscle synergies [175,176].

In non-stepping balance control, the counter-rotation strategy has been suggested to be more robust than the ankle strategy [177,178], and the use of counter-rotation strategies relative to the ankle increases with age and the magnitude of perturbations [52,179–181]. Older adults rely on the counter-rotation strategy at a lower level of challenge than younger adults, even during unperturbed balancing [52,180]. This presumably helps to secure robust balance control regardless of age-related sensory errors [55,182].

Balance training has been shown to result in altered muscle synergies and kinematics after a perturbation [183,184]. This may reflect improved feedback control but may also reflect improved feedforward control. Previously, we showed that training of older adults focusing on balance control on unstable surfaces, improved performance in perturbed and unperturbed balance tasks [185]. In addition, we found that the duration of co-contraction of muscles around the ankle increased, and we suggested that this may reflect an improved feedforward control strategy that contributed to performance improvements. Thus, training may have improved feedforward control resulting in less use of the counter-rotation strategy for balance recovery after a perturbation. However, in spite of the fact that the training program did not contain sudden, unpredictable perturbations, the challenging exercises used in training may also have improved feedback balance control, in which case one might expect the more effective counter-rotation strategy to be used more after training. In this study, we investigated the effects of training on kinematics and muscle synergies of balance recovery after perturbations in more detail to improve our understanding of training effects on feedback control of balance in older adults.

Methods

The data collection and training were described earlier [185], here we provide a brief summary. In this study twenty older adults (71.9±4.09 years old) participated. All participants provided written informed consent prior to participation, and the ethical review board of the Faculty of Behavioural and Movement Sciences, Vrije Universiteit Amsterdam, approved the experimental procedures (VCWE-2018-171).

Training consisted of balancing on balance boards and foam pads. The first training session was completed individually (30 minutes), and subsequently, a 3-weeks training program was completed in groups of 6-8 participants (45x3 minutes per week). We gradually increased the challenge of exercises by reducing hand support, moving from bipedal to unipedal stance, using more unstable support surfaces, and adding perturbations such as catching and throwing a ball and reducing visual input.

We assessed balance recovery with participants in unipedal stance on their dominant leg on a robot-controlled platform (HapticMaster, Motek, Amsterdam, the Netherlands). Participants performed 5 trials of a perturbed unipedal balance task, in which 12 random perturbations (6 medial and 6 lateral) were induced during 50-60 seconds. The platform rotated over a sagittal axis in the medial or lateral direction (amplitude of 8°) in random order. Participants were given two minutes rest between trials and a randomized 3-5 seconds rest period between perturbations within the trial. Participants were asked to fix their vision on a target in front of them. Full-body 3D kinematics were tracked by one Optotrak camera array (Northern Digital, Waterloo, Canada). Surface electromyography (EMG) data were recorded from nine unilateral muscles of the dominant leg: tibialis anterior (TA), vastus lateralis (VL), lateral gastrocnemius (GsL), soleus (SOL), peroneus longus (PL), rectus femoris (RF), biceps femoris (BF) and gluteus medius (GlM) and erector spinae (ES) muscles (TMSi, Twente, The Netherlands). We collected the data at baseline (Pre), after one training session (Post1), and after ten training sessions (Post2).

Data analysis

Sixty perturbations per participant per time-point (30 medial and 30 lateral) were used to calculate all variables.

Perturbation Onset

The onset of the perturbations was detected through the platform's rotation angle after synchronizing the platform, kinematics, and EMG data. Medial perturbations were defined when the platform started to rotate such that the big toe moved downward (eversion) and lateral when the big toe moved upward (inversion), and this was consistent for right- and left-leg dominant participants. A time window from 0.5 s before the onset of the perturbation to 2.5 s after the onset was selected for further analysis of all variables. For all variables 0.5 s baseline (from the start of the window until perturbation onset) was subtracted. Kinematics data were ensemble-averaged first over perturbations within a trial and then over trials per participant. The selected window was divided into three sub-windows; phase1 from perturbation onset to the maximum rotation angle of the platform, phase 2 (return to baseline) from maximum angle to 0-degree rotation angle of the platform and phase3 for 1 s after the platform returned to a 0-degree orientation.

Balance recovery, performance and strategy

We averaged the time series of center of mass displacement (CoM [m]), velocity (vCoM [m/s]) and acceleration (aCoM [m/s^2]) in the frontal plane over all trials at a given time-point per subject. We calculated the positive and negative areas under the center of mass acceleration curve as an indicator of balance performance[186]. Next, we calculated total body angular momentum [$kg.m^2/s$], its integral after division by the instantaneous moment of inertia to obtain a description of body orientation [degree], and the rate of change in total body angular momentum (time derivative of the total body angular momentum [$kg.m^2/s^2$]). We calculated the positive and negative areas under the curve of the rate of change of angular momentum as a second indicator of performance. The positive and negative areas were estimated separately for the three phases per direction of perturbation. The counter-rotation strategy is used when the rate of change in angular momentum accelerates the center of mass towards the base of support. Independent of this, angular momentum may be changed to regain upright body orientation. To assess how angular momentum changes were used, we compared the direction and timing of changes in the rate of change of angular momentum with CoM position and with body orientation.

Muscle synergies

EMG data were high-pass (35 Hz, bidirectional, 3rd order Butterworth) and notch filtered (50 Hz and its harmonics up to the Nyquist frequency, 1 Hz bandwidth, bidirectional, 1st order Butterworth). The filtered data were Hilbert transformed, rectified, and low-pass filtered (20 Hz, bidirectional, 3rd order Butterworth). Each rectified EMG signal was normalized to the maximum EMG value obtained over five perturbation trials per participant per time-point. The EMG data were down sampled to the sampling rate of the balance platform (100 Hz) to speed up the calculations. Subsequently, windows were selected from 0.5 s before until 2.5 s after the onset of perturbation. All windows for all participants and time-points were concatenated per perturbation direction. From these concatenated data, synergies were decomposed into a weighting matrix (spatial component) and activation profiles (temporal component) using non-negative matrix factorization (NNMF). Four synergies were extracted per perturbation direction, such that reconstructed EMG data accounted for a minimum of 90% of the variance in the EMG data[187]. Subsequently, we reconstructed the temporal activation profiles using pseudo-inverse multiplication of EMG data at the original sampling rate (2000 Hz) with the spatial components resulting from the previous step, per participant per time-point. The baseline values (mean over half a second before onset) were subtracted to focus on changes in muscle activation after perturbation.

We analyzed the activation profiles by estimating the magnitude of the activation and time to the peak activation for both the positive (excitation) and negative (inhibition) parts of the curve. The magnitude of the activation profile was calculated as the positive and negative area under the time dependent activation profiles separately for three phases. Time to the peak was estimated as the time that the maximum and minimum peak of an activation profile occurred in the selected window per synergy, per direction of perturbation. Magnitudes and time to the peak activation were averaged over 30 perturbations per participant and per time-point per direction of perturbation.

Statistics

One-way repeated-measures ANOVAs were used to identify the main effect of Time-point (Pre, Post1, Post2) on all kinematics and synergy variables per phase except for time to the peak. For time to the peak the statistical analysis was performed for the whole perturbation duration, as the peaks could have been shifted between phases after the training. Greenhouse-Geisser corrections were used when the assumption of sphericity was violated. In case of a significant effect of Time-point, post-hoc tests with Holm's correction for multiple comparisons (Pre-Post1 and Pre-Post2) were performed. In all statistical analyses, $\alpha = 0.05$ was used.

Results

Balance performance

In figure 4.1 and 4.2, center of mass displacement, velocity, and acceleration (left panels), as well as orientation, angular momentum, and rate of change of angular momentum (right panels) are displayed. The phases are color-coded. In phase 1, the initial change in angular momentum and center of mass acceleration are in line with the direct effects of the perturbations. However, corrective responses can be observed since acceleration and rate of change of angular momentum changed direction before the platform reached its maximum angle. In phase 2, corrective responses further counteracted the induced angular momentum. These responses did not correct the center of mass position, but rather corrected the upper body orientation to vertical. In phase 3, the platform stopped, and CoM and orientation cross the baseline and some overshoot in both occurs. In all three phases, but most obviously in phases 1 and 2, the sign of the rate of change in angular momentum would not result in accelerations that would correct CoM position as in the counter-rotation strategy, but instead corrected body orientation. This is illustrated in the drawings in figures 4.1 and 4.2. The left drawing illustrates the effect of the perturbation. The right figure illustrates the corrective response of the subject rotating the body in the opposite direction relative to the platform rotation, which induces an acceleration of the CoM in the direction of the perturbation.

The time series in figures 4.1 and 4.2 suggest that corrective responses were less pronounced after training, particularly after long-term training. This could be due to both improved feedforward and feedback control. Feedforward control may have affected the kinematics particularly in phase 1, where perturbation effects seemed somewhat smaller after training, especially visible after medial perturbations. In phase 2, after the training, corrective responses seemed attenuated, resulting in less overshoot in phase 3. Statistical analyses of these effects are reported below for lateral and medial perturbations separately.

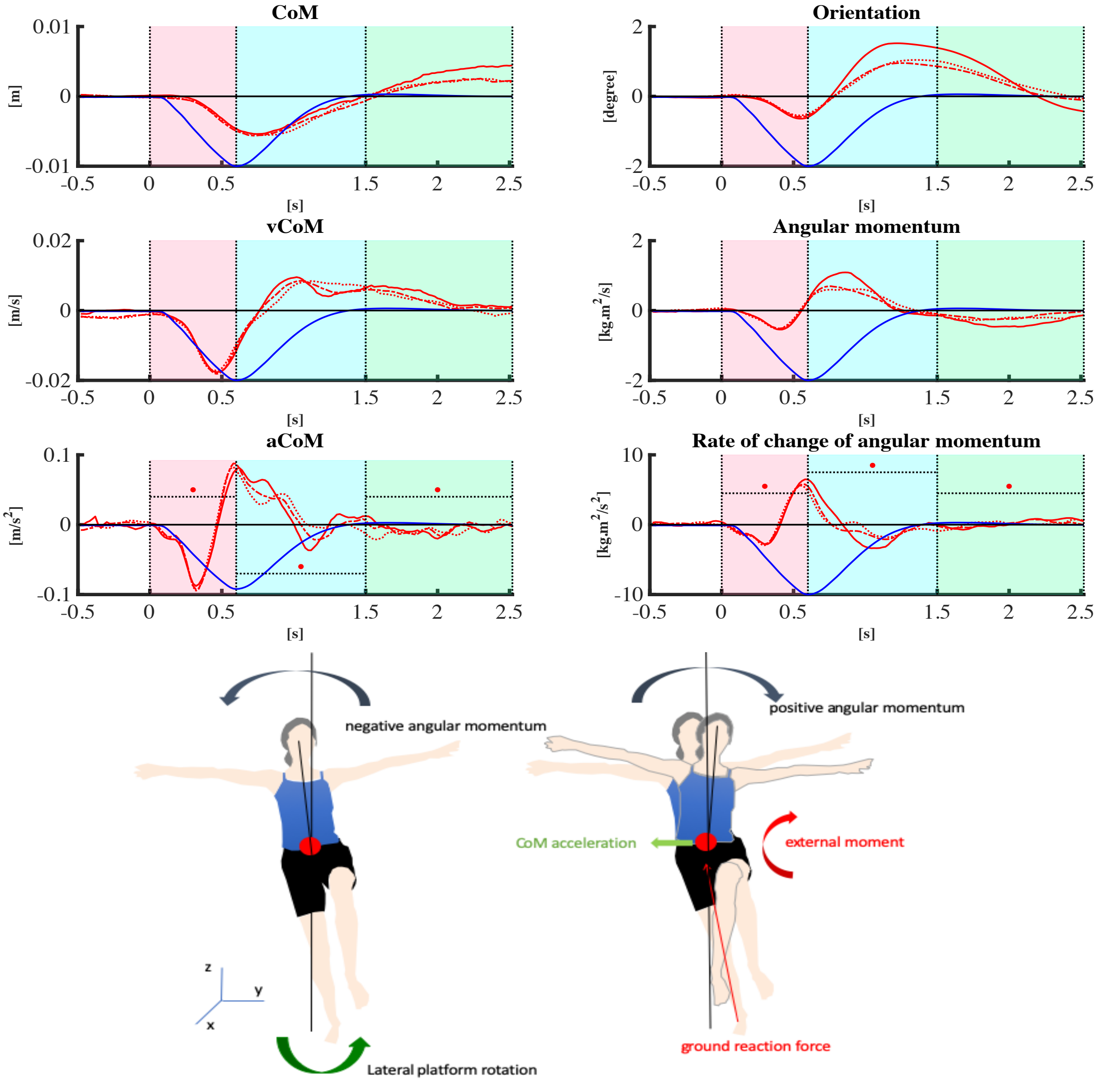

Figure 4.1: Linear kinematics (left panel) and rotational kinematics (right panel) are depicted for 0.5 s before onset to 2.5 s after onset of the lateral perturbations. Line types reflect Time-points: Pre (— solid line), Post1 (—• dash-dotted line), and Post2 (• dotted line). The red lines represent lateral perturbations; the blue line represents the rotation angle of the platform and is scaled per figure. Asterisks in 2 bottom subplots indicate a significant effect of training. The drawings illustrate the effect of the perturbation and the initial corrective response on angular momentum. The left drawing illustrates the initial, direct effect of the perturbation. The right figure illustrates the corrective response of the subject.

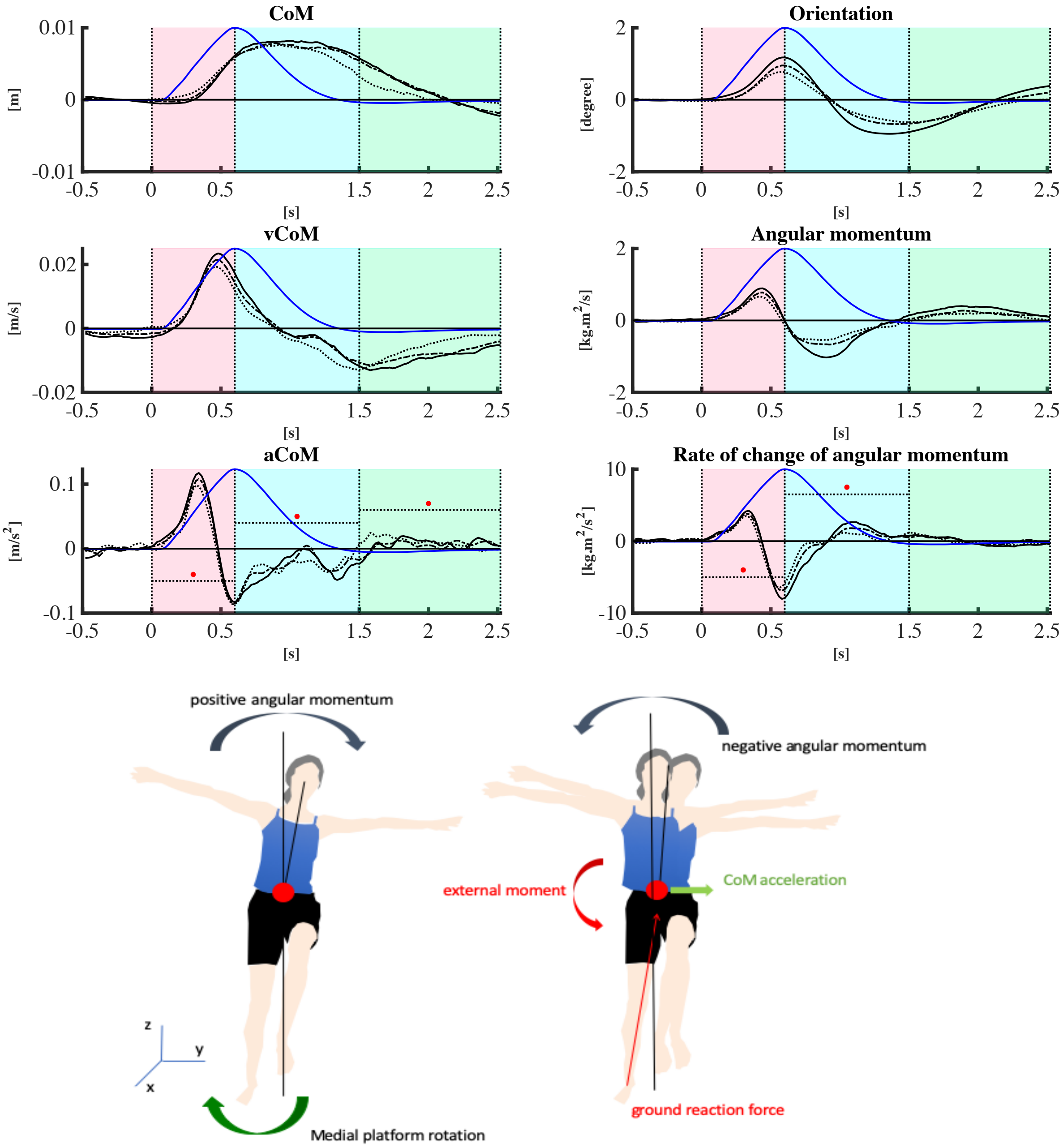

Figure 4.2: Linear kinematics (left panel) and rotational kinematics (right panel) are depicted for 0.5 s before onset to 2.5 s after onset of the medial perturbations. Lines reflect Time-points: Pre (— solid line), Post1 (—• dash-dotted line), and Post2 (• dotted line). The black lines represent the medial perturbations. The blue line represents the rotation angle of the platform and is scaled per figure. Asterisks in 2 bottom subplots indicate a significant effect of training. The drawings illustrate the effect of the perturbation and the initial corrective response on angular momentum. The left drawing illustrates the initial, direct effect of the perturbation. The right figure illustrates the corrective response of the subject.

Lateral perturbations

The negative area under the acceleration curve in phase 1, in the direction of the platform rotation, was affected by training ($F_{2,38}$ = 3.53, p = 0.039). Post-hoc testing showed that area under the acceleration curve did not change after short-term (p = 0.236), but decreased after long-term training (t = 2.63, p = 0.036; Figure 4.3.b, left panel). In phase 2, the positive area under the acceleration curve, in the direction of the platform rotation returning to horizontal, was also affected by training ($F_{2,38} = 7.46$, $p = 0.002$). Post-hoc testing showed that area under the acceleration curve decreased after both short- and long-term training ($t = 2.77$, $p = 0.017$; $t = 3.71$, $p = 0.002$, respectively; Figure 4.3.a, middle panel). Also, in phase 2, the negative area under the acceleration curve in the direction opposite to the perturbation angle was affected by training ($F_{2,38} = 3.90$, $p = 0.029$). Post-hoc testing showed that area under the acceleration curve did not change after short-term ($p = 0.092$) but decreased after long-term training ($t = 2.66$, $p = 0.034$; Figure 4.3.b, middle panel). In phase 3, the positive area under the acceleration curve was affected by training ($F_{2,38} = 9.24$, $p < 0.001$). Post-hoc testing showed that the area under the acceleration curve decreased after both short- and long-term training ($t = 3.14$, $p = 0.006$; $t = 4.11$, $p < 0.001$, respectively; Figure 4.3.a, right panel).

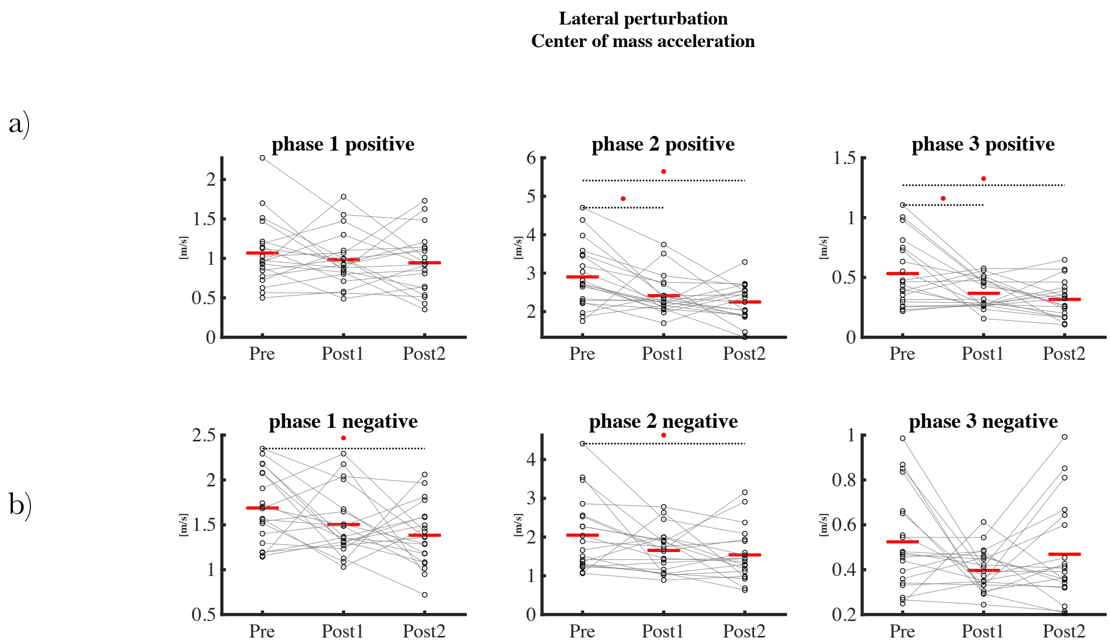

Figure 4.3: Area under center of mass acceleration curve after lateral perturbations at three time-points. Top panel a), represents the positive area, and bottom panel b) represents the negative area. Phase 1, 2 and 3 are shown in left, middle and right panel, respectively.

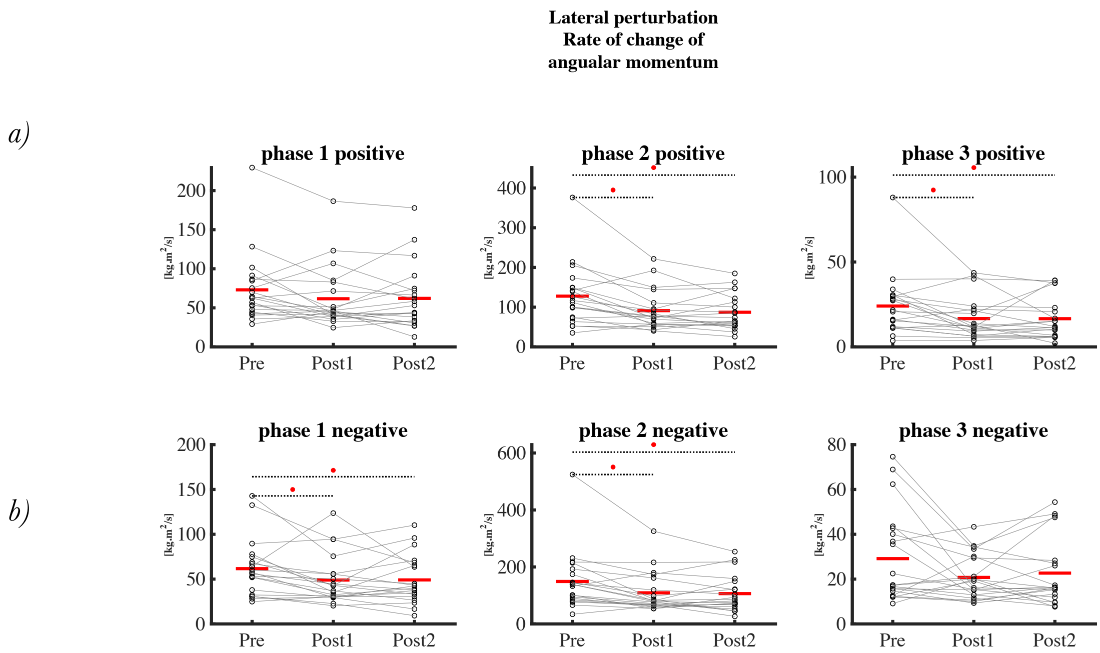

Figure 4.4: Area under the curve of the rate of change of angular momentum after lateral perturbations at three time-points. Top panel a), represents the positive area, and bottom panel b) represents the negative area. Phase 1, 2 and 3 are shown in left, middle and right panel, respectively.

In phase 1, the initial negative area under the rate of change of angular momentum curve, in the direction of the platform rotation, was affected by training ($F_{2,38} = 4.52$, $p = 0.017$). Post-hoc testing showed that area under the rate of change of angular momentum curve decreased after both short- and long-term training ($t = 2.62$, $p = 0.038$; $t = 2.59$, $p = 0.038$, respectively; Figure 4.4.b, left panel). In phase 2, the positive area under the rate of change of angular momentum curve, in the direction of the platform rotation returning to horizontal, was affected by training ($F_{1.47,28.09} = 11.34$, $p < 0.001$). Post-hoc testing showed that area under the rate of change of angular momentum curve decreased after both short- and long-term training ($t = 3.89$, $p < 0.001$; $t = 4.32$, $p < 0.001$, respectively; Figure 4.4.a, middle panel). Also, in phase 2, the negative area under the rate of change of angular momentum curve, opposite to platform rotation, was affected by training ($F_{1.32,25.16} = 7.68$, $p = 0.006$). Post-hoc testing showed that area under the rate of change of angular momentum curve decreased after both short- and long-term training ($t = 3.27$, $p = 0.005$; $t = 3.50$, $p = 0.004$, respectively; Figure 4.4.b, middle panel). In phase 3, the positive area under the rate of change of angular momentum curve was affected by training ($F_{2,38} = 4.87$, $p = 0.013$). Post-hoc testing showed that area under the rate of change of angular momentum curve decreased after both short- and long-term training ($t = 2.69$, $p = 0.03$; $t = 2.71$, $p = 0.03$, respectively; Figure 4.4.a, right panel).

Medial perturbations

In phase 1, the positive area under the acceleration curve, in the direction of the platform rotation, was affected by training ($F_{2,38} = 8.61$, $p < 0.001$). Post-hoc testing showed that area under the acceleration curve did not change after short-term ($p = 0.07$) but decreased after long-term training ($t = 4.14$, $p < 0.001$; Figure 4.5.a, left panel). In phase 2, the negative area under the acceleration curve, in the direction of the platform rotation back to horizontal, was affected by training ($F_{2,38} = 7.46$, $p = 0.002$). Post-hoc testing showed that area under the acceleration curve did not change after short-term ($p = 0.092$) but decreased after long-term training ($t = 2.72$, $p = 0.029$; Figure 4.5.b, middle panel). Also, in phase 2, the positive area under the acceleration curve, opposite to the direction of platform rotation, was also affected by training ($F_{2,38} = 3.78$, $p = 0.032$).

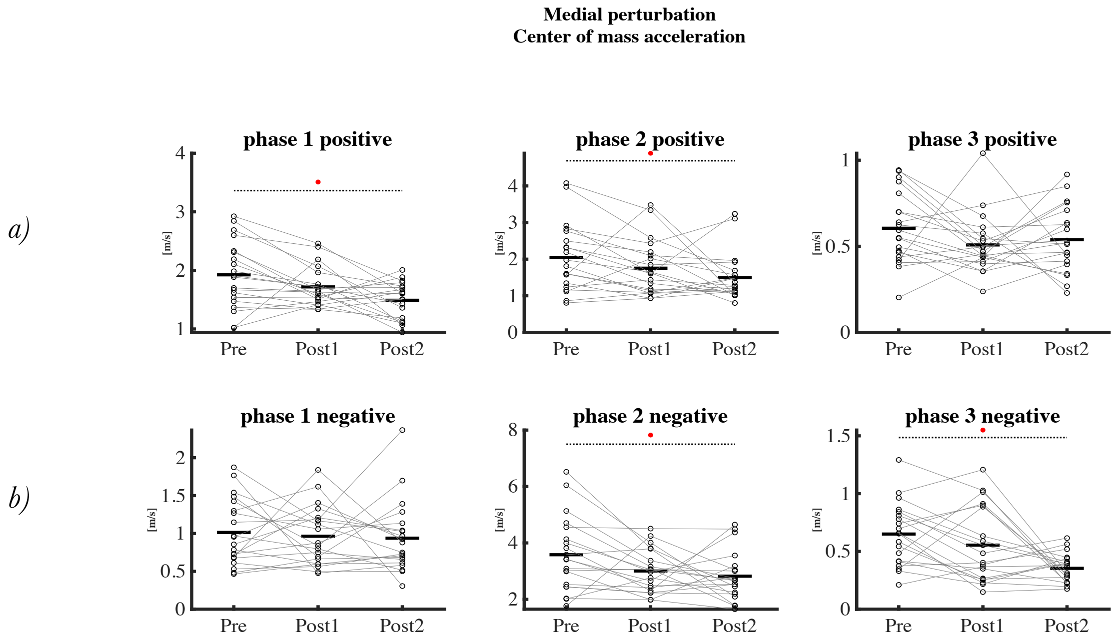

Figure 4.5: Area under the curve of the center of mass acceleration after medial perturbations at three time-point. Top panel a), represents the positive area, and bottom panel b) represents the negative area. Phase 1, 2 and 3 are shown in left, middle and right panel, respectively.

Post-hoc testing showed that the area under the acceleration curve did not change after short-term ($p = 0.301$) but decreased after long-term training ($t = 2.74$, $p = 0.027$; Figure 4.5.a, middle panel). In phase 3, the negative area under the acceleration curve was affected by training ($F_{2,38} = 8.63$, $p < 0.001$). Post-hoc testing showed that area under the acceleration curve of overshoot did not change after short-term ($p = 0.195$) but decreased after long-term training ($t = 4.072$, $p < 0.001$; Figure 4.5.b, right panel).

In phase 1, the initial, positive area under the rate of change of angular momentum curve, in the direction of the platform rotation, was affected by training ($F_{2,38}$ = 7.13, p = 0.002). Post-hoc testing showed that the area under the rate of change of angular momentum curve decreased after both short- and long-term training (t = 2.59, p = 0.027; t = 3.67, p = 0.002, respectively; Figure 4.6.a, middle panel).

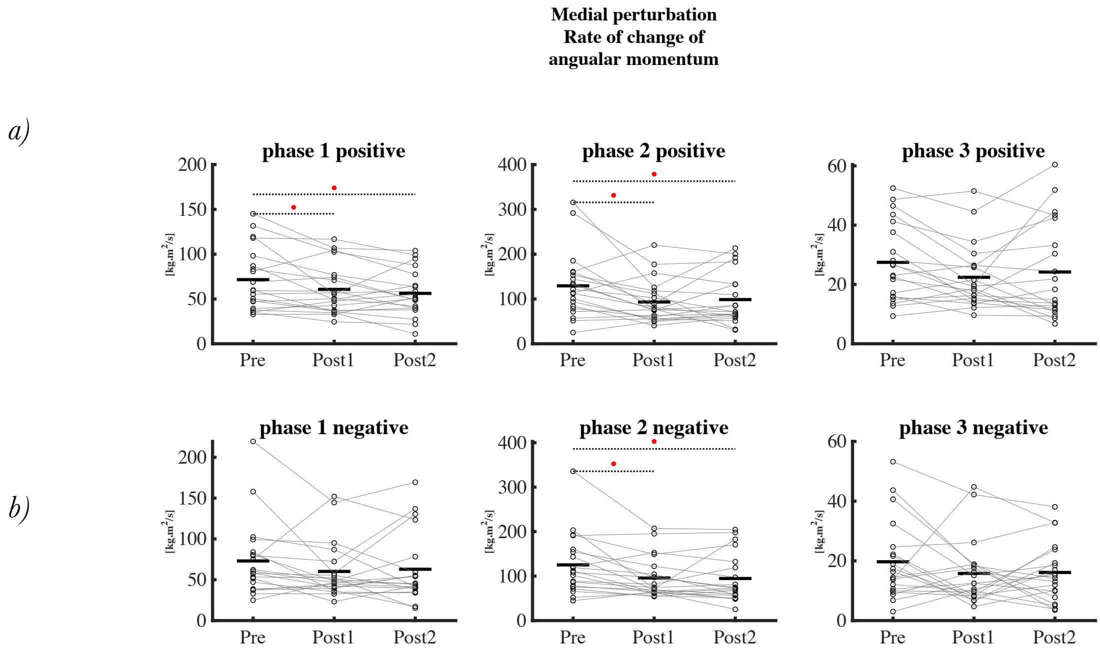

Figure 4.6: Area under the curve of the rate of change of angular momentum after medial perturbations at three time-points. Top panel a), represents the positive area, and bottom panel b) represents the negative area. Phase 1, 2 and 3 are shown in left, middle and right panel, respectively.

In phase 2, the negative area under the rate of change of angular momentum curve, in the direction of the platform rotation back to horizontal, was also affected by training ($F_{2,38}$ = 7.26, p = 0.002). Post-hoc testing showed that the area under the rate of change of angular momentum curve decreased after both short- and long-term training (t = 3.22, p = 0.005; t = 3.36, p = 0.005, respectively; Figure 4.6.b, middle panel). The later positive area under the rate of change of angular momentum curve in phase 2 was also affected by training ($F_{2,38}$ = 5.67, p = 0.007). Post-hoc testing showed that area under the rate of change of angular momentum curve decreased after both short- and long-term training (t = 3.12, p = 0.01; t = 2.65, p = 0.023, respectively; Figure 4.6.a, middle panel). In phase 3, no effects of training on the positive or negative area under the rate of change of angular momentum curve were found (p = 0.058, p= 0.298, respectively).

Muscle Synergies

The spatial components, i.e., the weighting factors of muscles per synergy, were largely similar between medial and lateral perturbations (Figure 4.7). The activation profiles in synergy 2 including lateral ankle muscles and in synergy 3 including frontal ankle muscles seemed to be mirrored between lateral and medial perturbations, and therefore might reflect the use of the ankle strategy for mediolateral stabilization. But interestingly, the initial responses in these synergies, which could reflect stretch responses of the ankle muscles, would aggravate the effect of the perturbation. Synergy 1 included GlM and RF showed a fairly similar activation profile for medial and lateral perturbations and may thus be less relevant for mediolateral stabilization. Synergy 4 included the erector spinae on the stance legs side, and was mainly active after medial perturbations and may be relevant for control of upper body orientation.

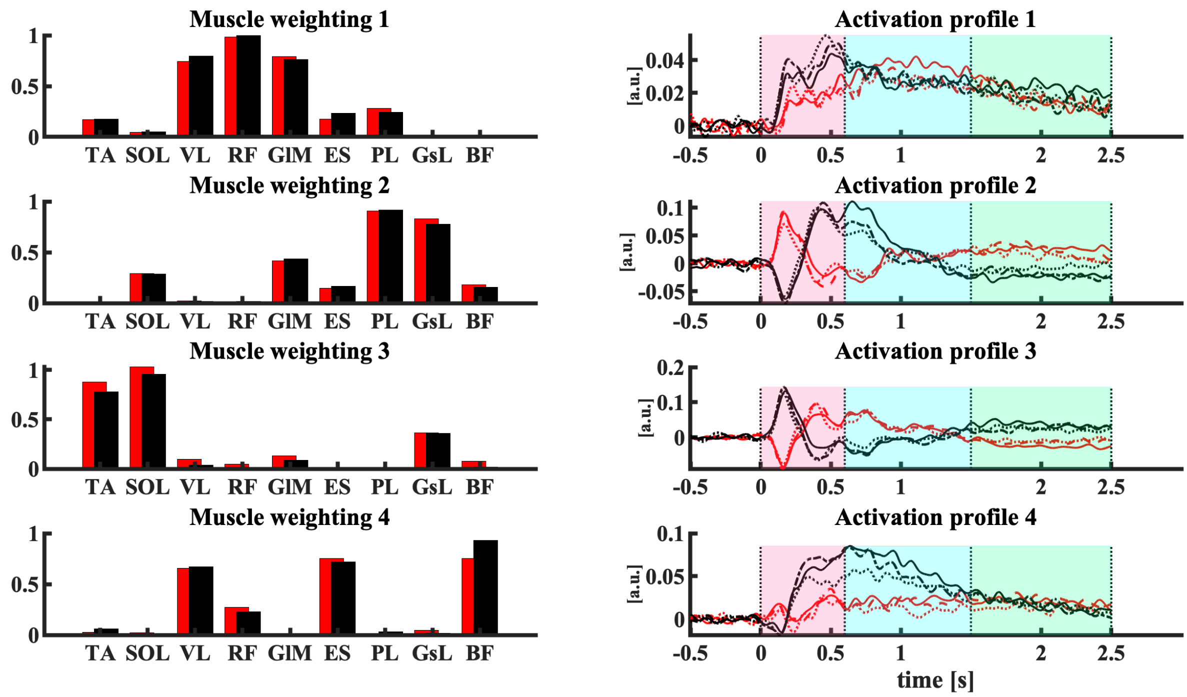

Figure 4.7: The muscle weighting and average activation profile shown after medial and lateral perturbations for three phases. Phase1 from perturbation onset to the maximum rotation angle of the platform, phase 2 (return to baseline) from maximum angle to 0-degree rotation angle of the platform and phase3 for 1 s after the platform returned to a 0-degree orientation. Lines reflect Time-points: Pre (— solid line), Post1 (—• dash-dotted line), and Post2 (• dotted line). The red color is assigned to lateral and the black color is assigned to medial perturbations. The baseline values in temporal activation in panel right were subtracted after the factorization to identify suppression and excitation relative to baseline values.

Lateral perturbations

In phase 1, no significant changes were observed after training.

In phase 2, although the activation profile was mainly above baseline (excitation), there was a significant effect of training on the negative area under the curve (inhibition) of synergy 1

($F_{2,38} = 3.62$, $p = 0.036$). However, post-hoc testing showed no significant changes after short- nor long-term training ($t=2.38$, $p = 0.066$; $t=2.27$, $p = 0.066$, respectively; Figure 4.8). Training also affected the negative area under the curve (inhibition) of synergy 4 in phase 2 ($F_{2,38} = 4.31$, $p = 0.02$), although the average activation in this phase was positive. Post-hoc testing showed that activation was not changed after short-term training, but was more inhibited after long-term training ($t=0.37$, $p = 0.70$; $t=2.71$, $p = 0.03$, respectively; Figure 4.8).

In phase 3, there was an effect of training on the positive area under the curve (excitation) of synergy 3 ($F_{2,38} = 3.67$, $p = 0.035$). Post-hoc testing showed that the activation did not significantly change after short-term but the positive area was larger after long-term training ($t = -2.08$, $p = 0.088$; $t = -2.54$, $p = 0.045$, respectively; Figure 4.8).

The time to the peak did not significantly change after the training in any of the synergies.

Lateral perturbation

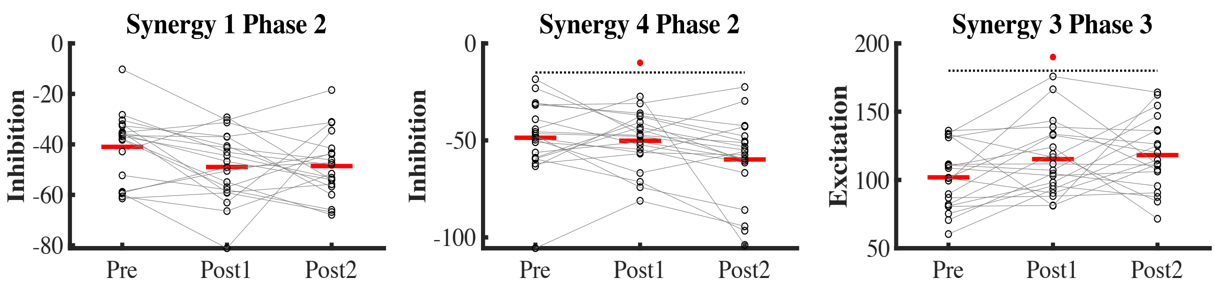

Figure 4.8: Area under the curve of activation profiles in selected phases after lateral perturbations at three time-points.

Medial perturbations

In phase 1, there was an effect of training on the negative area under the curve (inhibition) of synergy 2 ($F_{2,38} = 3.52$, $p = 0.039$) which was generally less strong after training, although post-hoc testing showed no significant differences after short-term or long-term training ($t= -0.38$, $p = 0.70$; $t= -2.46$, $p = 0.055$, respectively; Figure 4.9). Training increased the later negative area under the curve of synergy 3 in phase 1 ($F_{2,38} = 7.53$, $p = 0.002$). Post-hoc testing showed that activation was more inhibited after both short- and long-term training ($t=2.71$, $p = 0.02$; $t=3.76$, $p = 0.002$, respectively; Figure 4.9). Training also affected the initial negative area under the curve of synergy 4 in phase 1 ($F_{2,38} = 4.99$, $p = 0.012$). Post-hoc testing showed that activation did not change after short-term training, but was more inhibited after long-term training ($t = 0.811$, $p = 0.422$; $t = 3.05$, $p = 0.012$, respectively; Figure 4.9).

In phase 2, there was an effect of training on the excitation of synergy 2 ($F_{2,38} = 3.37$, $p = 0.045$). Post-hoc testing showed that the activation profile did not change after short- nor after long-term training (t= 2.46, p = 0.055; t=1.94, p = 0.119, respectively; Figure 4.9).
In phase 3, training affected the inhibition of synergy 2 in overshoot ($F_{2,38} = 3.96$, $p = 0.027$). Post-hoc testing showed that the activation profile did not change after short-term but was less inhibited after the long-term training (t= - 0.86, p = 0.39; t= - 2.75, p = 0.027, respectively; Figure 4.9). The inhibition of synergy 1 in overshoot, phase 3, was also affected by training ($F_{2,38} = 3.36$, $p = 0.045$). Post-hoc showed it was more inhibited after short-term but back to baseline after long-term training (t = 2.53, p = 0.047; t =1.74, p = 0.177, respectively; Figure 4.9).

The time to the peak in synergy 2 after medial perturbation was affected by training ($F_{2,38} = 6.40$, $p = 0.004$). Post hoc showed that time to the peak did not change after short- but was delayed after long-term training (t=-2.10, p = 0.085; t=-3.559, p = 0.003, respectively; Figure 4.9).

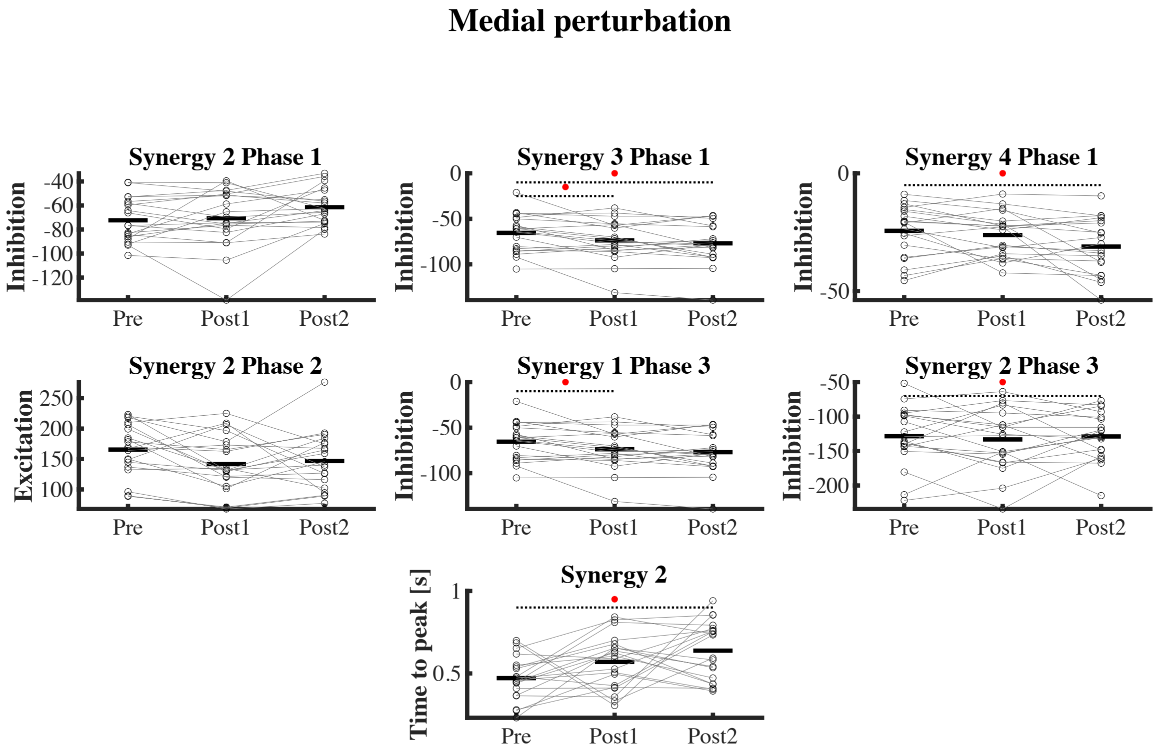

Figure 4.9: Area under the curve of activation profiles in selected phases after medial perturbations at three time-points.

Discussion

Aging comes with challenges to recover from perturbed balance. An effective balance training could reduce such challenges. Previously we reported that training decreased mean absolute center of mass velocity and increased ankle muscle co-contraction in perturbed unipedal balancing [185]. We suggested that increased co-contraction might compensate for the age-related deficits in sensory-motor control and as such reflect improved feedforward balance control. Yet, feedforward control is not always sufficient. Moreover, feedforward control in the form of sustained co-contraction requires energy and hence could cause fatigue. In our experiment, participants were expecting a perturbation, but were unaware of its timing and direction, allowing some, but limited feedforward control. When balance was perturbed, consistent responses in muscle activations were observed, indicating that feedback control was still used to regain balance. After training, changes in feedback control and smaller corrective responses to reorient the upper body to the upright position were observed.

Medial and lateral platform perturbations caused corresponding medial and lateral accelerations of the CoM as well as a change in angular momentum in the direction of platform rotation. Subsequent changes in angular momentum did not contribute to moving the center of mass back to the baseline position which indicates that the counter-rotation strategy was not used by our participants. Thus, the rate of change in angular momentum was used to re-orient the body rather than to shift the center of mass position over the base of support. This reorientation of the body was better tuned after training, i.e., the corrective change in angular momentum had a smaller area under the rate of change of angular momentum curve, resulting in less overshoot. This also contributed to better control of the CoM as the adverse effect on CoM acceleration would be smaller. These findings emphasize that balance control, conceptualized as control over CoM position relative to the base of support, is constrained by control of body orientation. While the CoM could be maintained over the base of support with opposite orientations of the upper and lower body oriented, a vertical orientation of both segments seems to be preferred and would of course be less demanding.

Reactive balance control improved after training, as shown by decreased amplitudes of the center of mass acceleration and rate of change of angular momentum. The improvement in center of mass acceleration and in rate of change of angular momentum in phase 1 might partially be caused by better feedforward control or improved reflex-based activity immediately following the perturbation and partially by improved feedback control in generating the corrective response at the end of the phase 1. The significant improvement in balance performance of phase 2 after perturbations indicates that the feedback control of balance improved more notably after long-term training. The improvements in phase 3 are likely an effect of better tuned responses in phase 2, resulting in less overshoot, but could also be due to higher co-contraction leading to a quicker damping of oscillations after the perturbations.

For lateral perturbations, training caused changes in the synergies in phases 2 and 3. After training and most notably after long-term training, participants inhibited synergy 4, comprising ES, VL and BF muscles, more in phase 2. This may have reduced the overshoot of CoM movement augmented by an increase in co-contraction of TA and SOL as evidenced by enhanced excitation of synergy 3 in phase 3.

For medial perturbations, training caused less inhibition in phase 1 of synergy 2 including the PL and GsL muscles. Since this initial inhibition may aggravate the perturbation, the training adaptation is likely beneficial. In the same phase the inhibition of synergy 4 including the ES, VL and BF muscles became more pronounced. The inhibition of the ES may have limited upper body rotation towards medial. Later in phase 1, synergy 3 including the TA and SOL was more inhibited and this inhibition likely helped balance recovery by the simultaneous excitation of synergy 2. In phase 2, synergy 2 was less activated after training, which may have reduced the overshoot of CoM movement, which in turn could explain the decreased inhibition of this synergy in phase 3.

Conclusion

We investigated the effect of balance training on feedback control after expected but unpredictable balance perturbations in older adults. Our results indicate that balance training improves performance and improves the corrective responses after a perturbation. The improvement was observed in reduced amplitudes of the rate of change of angular momentum and center of mass acceleration. The rate of change of angular momentum did not correct the center of mass position, as we expected from the definition of the counter-rotation strategy, but reoriented the body to the vertical. These kinematic changes appeared to be linked to altered temporal activation of muscles grouped in ankle and upper body synergies.

Acknowledgments

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Chapter 5

Neuromuscular control of gait stability in older adults is adapted to environmental demands but not improved after standing balance training

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Abstract

Balance training aims to improve balance and transfer acquired skills to real-life tasks and conditions. How older adults adapt gait control to different conditions, and whether these adaptations are altered by balance training remains unclear. We investigated adaptations in neuromuscular control of gait in twenty-two older adults (72.6 ± 4.2 years) between normal (NW) and narrow-base walking (NBW), and the effects of a standing balance training program shown to enhance unipedal balance control in the same participants. At baseline, after one session and after 3-weeks of training, kinematics and EMG of NW and NBW on a treadmill were measured. Gait parameters and temporal activation profiles of five synergies extracted from 11 muscles were compared between time-points and gait conditions. No effects of balance training or interactions between training and walking condition on gait parameters or synergies were found. Trunk center of mass (CoM) displacement and velocity (vCoM), and the local divergence exponent (LDE), were lower in NBW compared to NW. For synergies associated with stance of the non-dominant leg and weight acceptance of the dominant leg, full width at half maximum (FWHM) of the activation profiles was smaller in NBW compared to NW. For the synergy associated with non-dominant heel strike, FWHM was greater in NBW compared to NW. The Center of Activation (CoA) of the activation profile associated with dominant leg stance occurred earlier in NBW compared to NW. CoAs of activation profile associated with non-dominant stance and non-dominant and dominant heel strikes were delayed in NBW compared to NW. The adaptations of synergies to NBW can be interpreted as related to a more cautious weight transfer to the new stance leg and enhanced control over CoM movement in the stance phase. However, control of mediolateral gait stability and these adaptations were not affected by balance training.

Keywords: *Balance training, postural balance, aging, skill transfer, gait control, narrow-base walking, muscle synergy*

Introduction

Falls in older adults mostly occur during walking [188]. Therefore, skills acquired during standing balance training should transfer to gait and improve gait stability [189]. While on one hand effects of balance training have been described as task specific [158], on the other hand, transfer from standing balance training to gait stability has been suggested by improved clinical balance scores and gait parameters [190,191]. Consequently, the existence of skill transfer from standing balance training as well as the mechanisms underlying such transfer, if present, are insufficiently clear.

Increased variability and decreased local dynamic stability of steady-state gait were shown to be associated with a history of falls in older adults [192]. From a mechanical perspective, larger mediolateral center of mass excursions and velocities would be expected to cause an increased fall risk [193] and both these parameters as well as their variability are larger in older than young adults [194]. When facing environmental challenges, such as when forced to walk with a narrow step width, individuals need to adapt their gait. Older adults show more pronounced adaptations to narrow-base walking compared to young adults [194], possibly because they are more cautious in the presence of postural threats [195]. Transfer of standing balance training to gait would be expected to result in increased gait stability, decreased CoM displacement and velocity, and decreased CoM displacement variability. In addition, an interaction between training and stabilizing demands may be expected. Increased confidence after training may result in less adaptation to a challenging condition. On the other hand, balance training may enhance the ability to adapt to challenging conditions.

The central nervous system is thought to simplify movement by activating muscles in groups, called muscle synergies, with the combination of synergies shaping the overall motor output [37,196]. Muscle synergies consist of time-dependent patterns (activation profiles) and time-independent factors (muscle weightings). Human gait has been described with four to eight muscle synergies [197–199] and reactive balance control was found to have four shared synergies with walking [199], which could be important for transfer from balance training to gait. Due to aging and changes in sensory and motor organs, adapted synergies are likely required to maintain motor performance [200,201]. Synergy analyses of gait revealed either fewer synergies in older adults than in young adults [202] or no differences [203]. Motor adaptation is assumed to result from altering synergies in response to task and environmental demands [40,204]. For example, widened activation profiles appear to be used to increase the robustness of gait in the presence of unstable conditions or unpredictable perturbations [40,41]. Long-term balance training might alter synergies in gait, and adaptation of synergies to task demands as has been shown in dancers [75,205] to achieve the alterations in CoM kinematics.

We investigated the adaptations in neuromuscular control of gait in older adults between normal and narrow-base walking, and the effect of short- and long-term standing balance training on this. To this aim, we used data from a previous study on standing balance training, from which we previously reported positive effects of training on standing balance robustness and performance, both after a single training session and after three weeks of training [206]. Here, we evaluate skill transfer to normal walking and narrow-base walking on a virtual beam, both on a treadmill. We used foot placement error to assess performance of narrow-base walking [207]. We focused on mediolateral balance control, as larger mediolateral instability has been shown to be associated with falls in older adults [208,209] and beam walking challenges mediolateral stability. We calculated the CoM displacement and CoM displacement variability, CoM velocity and the LDE as measures of gait stability and extracted muscle synergies to characterize effects on the neuromuscular control of gait and of adaptations to narrow-base walking.

Methods

The methods described here in part overlap with our previous paper [206], as data were obtained in the same cohort.

Participants

Twenty-two older (72.6 ± 4.2 years old; mean ± SD, 11 females) healthy volunteers participated in this study. Participants were recruited through a radio announcement, contacting older adults who previously participated in our research, flyers and information meetings. Individuals with obesity (BMI > 30), cognitive impairment (MMSE<24), peripheral neuropathy, a history of neurological or orthopedic impairment, use of medication that may negatively affect balance, inability to walk for 4 minutes without aid, and performing sports with balance training as an explicit component (e.g., Yoga or Pilates) were excluded. All participants provided written informed consent before participation and the procedures were approved by the ethical review board of the Faculty of Behavioural & Movement Sciences, VU Amsterdam (VCWE-2018-171).

Experimental procedures

Participants completed an initial measurement to determine baseline values (Pre), a single-session balance training (30-minutes), a second measurement (Post1) to compare to baseline to assess short-term training effects, a 3-week balance training program (9 sessions x 45 minutes training), and a third measurement (Post2) to compare to baseline to assess of long-term training effects (Figure 5.1).

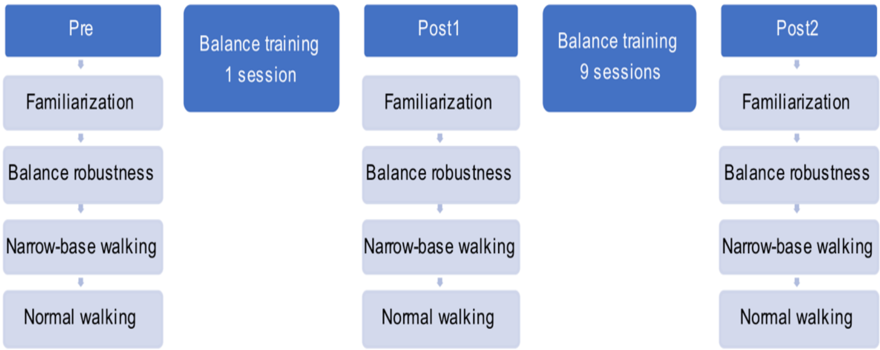

Figure 5.1: Block diagram of the study; training and gait assessment.

The measurements consisted of one experimental condition on a robot-controlled platform (balance robustness) and two experimental conditions performed on a treadmill: virtual-narrow-base walking (Figure 5.2) and normal walking.

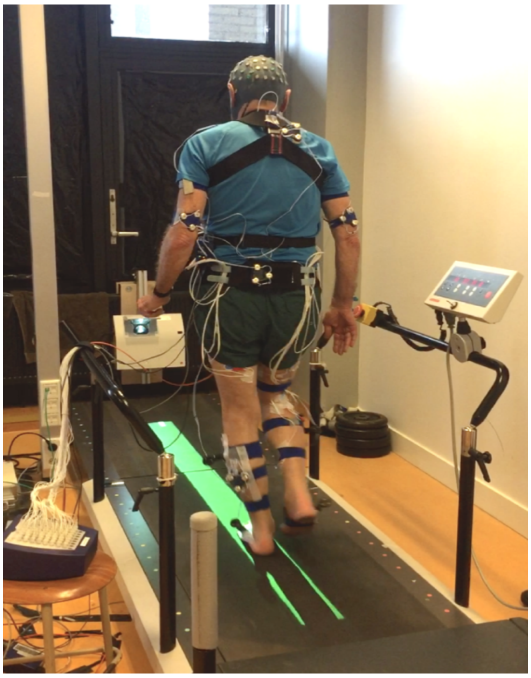

Figure 5.2: Narrow-base walking on a treadmill. Participant is wearing the EEG cap and in future work we will analyze the EEG data collected, to investigate changes at the supraspinal level.

The training sessions consisted of exercises solely focused on unipedal balancing with blocks of 40-60 second exercises in which balance was challenged by different surface conditions, static vs dynamic conditions, perturbations, and dual tasking (e.g. catching, throwing and passing a ball) [210]. Participants performed the exercises in a group of two (except for the first, individual session) and always under supervision of the physiotherapist in our research team.

Instrumentation and data acquisition

Balance robustness and performance were evaluated using a custom-made balance platform controlled by a robot arm (HapticMaster, Motek, Amsterdam, the Netherlands) and results were reported previously [206]. To quantify transfer to gait, participants were instructed to walk for 4.5 minutes at a constant speed of 3.5 km/h on a treadmill with an embedded force plate. For safety reasons, handrails were installed on the either side of the treadmill, and an emergency stop button was placed within easy reach (MotekForcelink, Amsterdam, the Netherlands). We assessed walking in two conditions, normal walking and narrow-base walking, in a randomized order, with a minimum of 2 minutes seated rest in between conditions. In narrow-base walking, participants were instructed to placing their entire foot inside the beam as accurately as possible over a green light-beam path (12 cm width) projected in the middle of the treadmill (Bonte Technology/ForceLink, Culemborg, The Netherlands) [207].

Kinematics data were obtained by two Optotrak 3020 camera arrays at 50 Hz (Northern Digital, Waterloo, Canada). 10 active marker clusters (3 markers each) were placed on the posterior surface of the thorax (1), pelvis (1), arms (2), calves (4), and feet (2) (Figure 5.2). Positions of anatomical landmarks were digitized by a 4-marker probe and a full-body 3D-kinematics model of the participant was formed relating clusters to the neighboring landmarks [156]. The position of the foot segments was obtained through cluster markers on both feet, digitizing the medial and lateral aspects of the calcaneus, and the heads of metatarsals one and five [207]. Additionally, to calculate the foot placement error in narrow-base walking, position and orientation of the projected beam was determined by digitizing the four outer bounds of the beam on the treadmill.

Surface electromyography (EMG) data were recorded from 11 muscles; 5 unilateral muscles of the dominant leg: tibialis anterior (TAD), vastus lateralis (VLD), lateral gastrocnemius (GLD), soleus (SOD), peroneus longus (PLD) and, 6 bilateral muscles: rectus femoris (RFD, RFN), biceps femoris (BFD, BFN) and gluteus medius (GMD, GMN) muscles. Bipolar electrodes were placed in accordance with SENIAM recommendations [95]. EMG data were sampled at a rate of 2000 Hz and amplified using a 16-channel TMSi Porti system (TMSi, Twente, The Netherlands). The dominant leg was the leg preferred for single-leg stance. Focus was on this leg, because we extensively assessed unipedal balance control on this leg as reported earlier [206].

Data analysis

Gait events

The first 30 seconds of all gait trials were removed, to discard the habituation phase. Heel-strikes were detected through a peak detection algorithm based on the center of pressure [211]. This algorithm proved to be precise when the center of pressure moved in a butterfly pattern. However, for narrow-base walking, the feet share a common area in the middle of the treadmill, therefore, identification of which leg touched the surface was problematic. Hence, heel-strikes were detected based on the center of pressure peak detection, but the associated leg was identified based on kinematic data of the foot marker. 160 strides per participant per condition were used to calculate all gait variables (i.e. stability variables and muscle synergies).

Gait stability

To evaluate gait performance, foot placement errors were determined as the mean mediolateral distance of the furthest edge of the foot from the edge of the beam. If the foot was within the beam the error equals zero.

The trajectory of the center of mass (CoM) of the trunk was estimated from mediolateral trunk movement [212,213]. As gait stability variables, we calculated mean and standard deviation of the peak-to-peak mediolateral trunk CoM displacement and mean of CoM velocity per stride. In addition, local dynamic stability was evaluated using the local divergence exponent, LDE, based on Rosenstein's algorithm [214,215]. We used the time normalized time-series (i.e. 160 strides of data were time normalized to 16000 samples, preserving between stride variability) of trunk vCoM to reconstruct a state space with 5 embedding dimensions at 10 samples time delay [213]. The divergence for each point and its nearest neighbor was calculated and the LDE was determined by a linear fit over half a stride to the averaged log transformed divergence.

Muscle synergies

EMG data were high-pass (50 Hz, bidirectional, 4th order Butterworth) [40] and notch filtered (50 Hz and its harmonics up to the Nyquist frequency, 1 Hz bandwidth, bidirectional, 1st order Butterworth). The filtered data were Hilbert transformed, rectified and low-pass filtered (10 Hz, bidirectional, 2nd order Butterworth). Each channel was normalized to the maximum activation obtained for an individual per measurement point per trial. Synergies were extracted from 11 muscles using non-negative matrix factorization. Five synergies were extracted from the whole dataset, to account for a minimum of 85% of the variance in the EMG data (Figure 5.6). It has been shown that perturbations during walking change the temporal activation profiles as compared to normal walking, while muscle weightings are preserved [216]. Therefore, in the current study we fixed muscle weightings between conditions and time-points. These muscle weightings were extracted from the concatenated EMG data of both conditions at all time-points. This allowed for objective comparison of synergy activation profiles between normal and narrow-base walking and between time-points. Consequently, the time-normalized EMG data of the muscles $E_{11 \times (2 \times 100 \times 160)}$, was factorized to two matrices: time-invariant muscle weightings, $W_{11 \times 5}$, and temporal activation profiles of the factorization, $A_{5 \times (2 \times 100 \times 160)}$, where 11 was the number of muscles, 5 the number of synergies, 2 the number of conditions, 100 the number of samples in each stride and 160 the number of strides. Afterwards, we reconstructed the temporal activation profiles using pseudo-inverse multiplication, for the comparison of activation profiles between conditions and time-points.

To compare activation profiles, we evaluated the full width at half maximum, FWHM, per stride for each activation profile (defined as the number of data points above the half maximum of activation profile, after subtracting the minimum activation [217]). In addition, we evaluated the center of activity, CoA, per stride defined as the angle of the vector that points to the center of mass in the activation profile transformed to polar coordinates [40,218]. FWHM and CoA were averaged over 160 strides per participant per condition. For CoA data, circular averaging was used.

Statistics

Effects of time-point (Pre, Post1, Post2) on foot placement errors were tested using a one-way repeated measures ANOVA. Post-hoc comparisons (paired sample t-tests), with Holm's correction for multiple comparisons were performed to investigate the effect of short- and long-term training (Pre vs Post1 and Pre vs Post2, respectively).

Two-way repeated-measures ANOVAs were used to identify main effects of time-point (Pre, Post1, Post2) and condition (normal and narrow-base walking) on trunk kinematics CoM displacement, CoM displacement variability, vCoM and LDE, as well as, on the FWHM. When the assumption of sphericity was violated, the Greenhouse-Geisser method was used. In case of a significant effect of time-point, or an interaction of time-point x condition, post hoc tests with Holm's correction for multiple comparisons were performed. To identify effects on CoA, parametric two-way ANOVA for circular data was used using the Circular Statistic MATLAB toolbox [219]. In all statistical analyses $\alpha = 0.05$ was used.

Results

One participant was not able to perform the treadmill walking trials for the full duration and data for this participant were excluded.

Gait performance

In contrast with robustness and performance in unipedal balancing [206], performance in narrow-base walking, as reflected in foot placement errors, did not improve as a result of training ($F_{1.267,25.347} = 0.31$, $p = 0.63$; Figure 5.3).

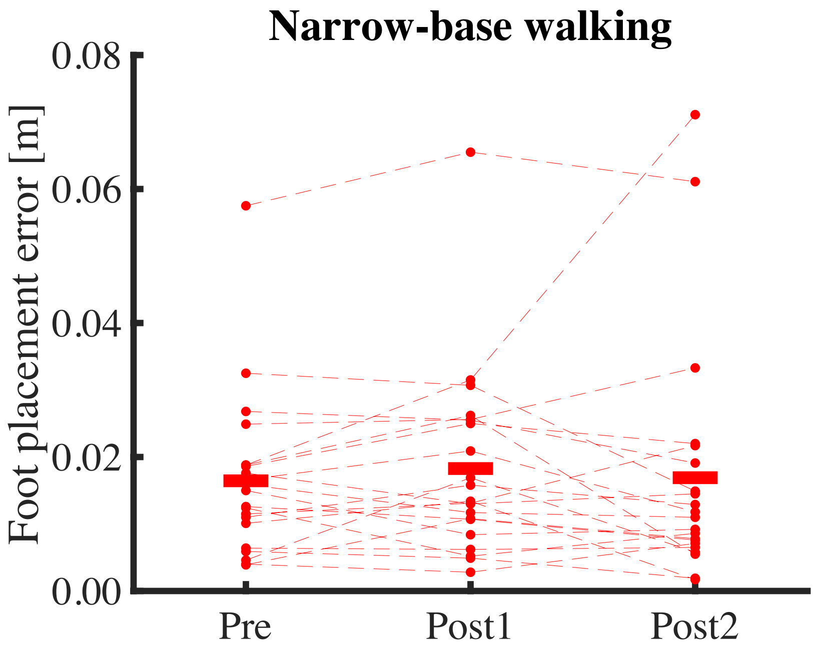

Figure 5.3: Foot placement error in narrow-base walking at time-points Pre, Post1, and Post2. Thin lines represent individual subject data. Red horizontal lines indicate means over subjects.

Training did also not significantly affect CoM displacement, CoM displacement variability, and vCoM ($F_{2,40} = 2.729$, $p = 0.082$; $F_{2,40} = 0.469$, $p = 0.628$; $F_{2,40} = 2.024$, $p = 0.145$). Condition significantly affected all three variables, with lower CoM and vCoM ($F_{1,20} = 96.007$, $p < 0.001$; $F_{1,20} = 168.26$, $p < 0.001$, respectively, Figure 5.4), but larger CoM variability ($F_{1,20} = 4.678$, $p = 0.042$), in narrow-base compared to normal walking. No significant interactions of time-point x condition were found ($p > 0.05$).

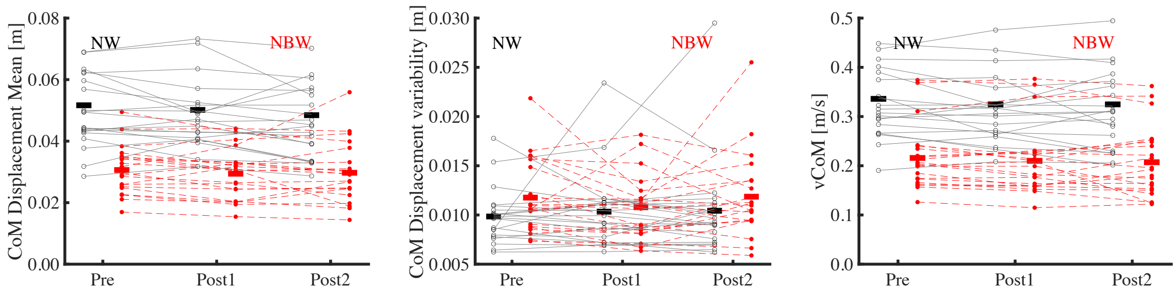

Figure 5.4: Mediolateral center of mass displacement, variability, and center of mass velocity in narrow-base and normal walking at time-points Pre, Post1, and Post2. Thin lines represent individual subject data. Thick horizontal lines indicate means over subjects. Black; normal walking, red; narrow-base walking.

Training did not significantly affect LDE ($F_{2,40} = 0.205$, $p = 0.814$), but condition did, with lower values in narrow-base compared to normal walking ($F_{1,20} = 26.223$, $p < 0.001$, Figure 5.5). No significant interaction of time-point x condition was found ($F_{1.3,24.699} = 3.112$, $p = 0.078$).

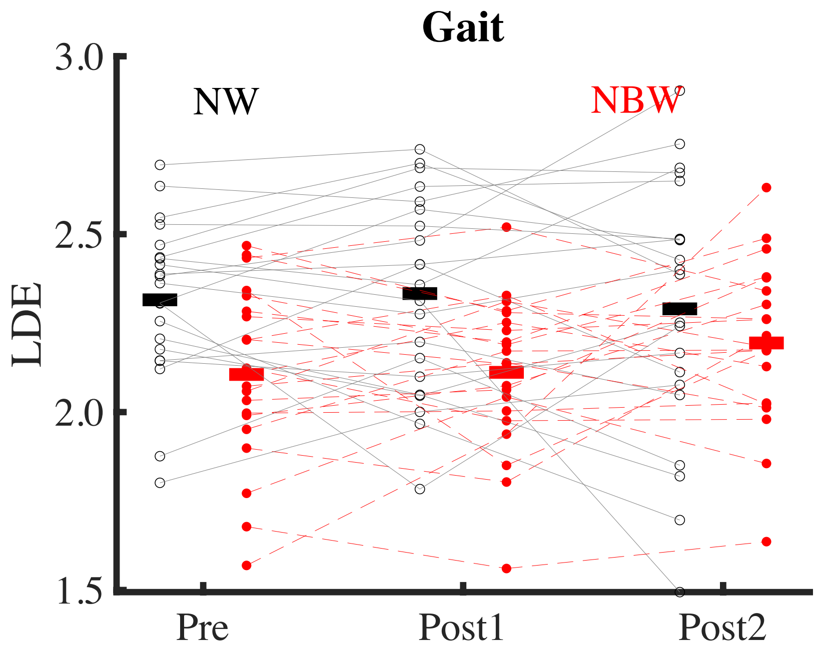

Figure 5.5: Local divergence exponents in narrow-base and normal walking at time-points Pre, Post1, and Post2. Thin lines represent individual subject data. Thick horizontal lines indicate means over subjects. Black; normal walking, red; narrow-base walking.

Muscle synergies

Five muscle synergies were extracted with a fixed muscle weighting matrix H (Figure 5.6) and activation profiles per individual per condition and time-point (Figure 5.7). This accounted for 87±2% of the variance in the EMG data.

Based on muscle weightings and activation profiles, the first synergy appeared to be functionally relevant in the stance phase of the dominant leg, with major involvement of soleus and gastrocnemius lateralis. The second synergy appeared to be related to the weight acceptance phase of the dominant leg, where the quadriceps (vastus lateralis, rectus femoris) muscles were mostly engaged. The third synergy resembled partial mirror images of synergies 1 and 2 for the non-dominant leg, but differed due to the fact that only a subset of muscles was measured. It was mainly active in the non-dominant leg's stance phase, with major involvement of gluteus medius and rectus femoris. It lacks muscle activation related to push-off (represented in synergy 1), because lower leg muscles were not measured and represented thigh muscle activity related to weight acceptance (represented in synergy 2). The fourth synergy appeared to anticipate dominant leg heel-strike with engagement mostly of the dominant leg's biceps femoris. Finally, the fifth synergy appeared to be the mirror image of the fourth synergy, with pronounced engagement of the biceps femoris of the non-dominant leg

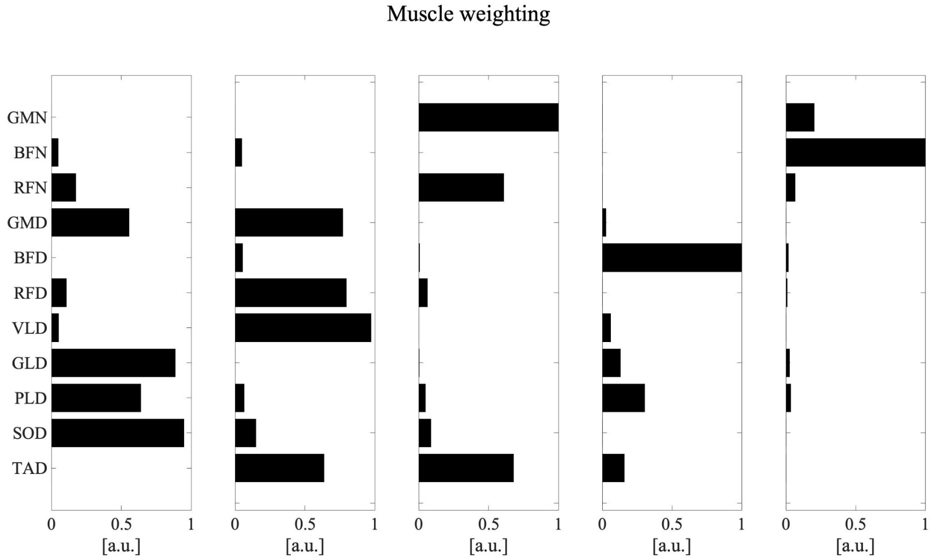

Figure 5.6: Time-invariant muscle weightings of synergies extracted from concatenated data, over all individuals, conditions and time-points. Muscles monitored unilaterally on the dominant side (D): tibialis anterior (TA), vastus lateralis (VL), lateral gastrocnemius (GLD, soleus (SO), peroneus longus (PLD), and muscle collected on the dominant (D) and non-dominant side (N): rectus femoris (RFD, RFN), biceps femoris (BFD, BFN), and gluteus medius (GMD, GMN) muscles.

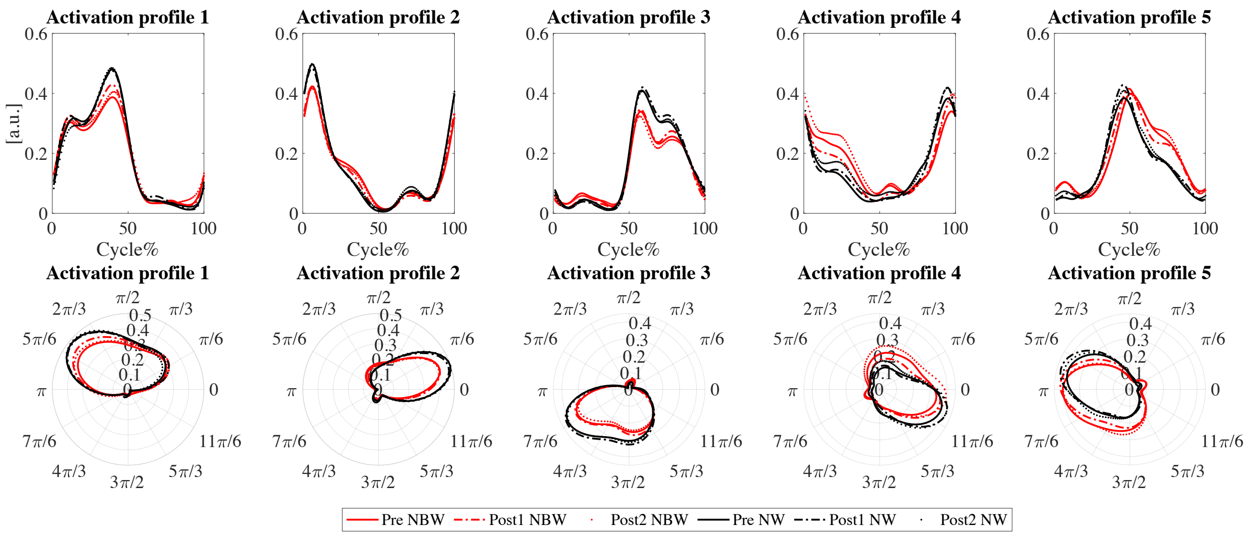

Figure 5.7. Activation profiles of the extracted synergies as time series and in polar coordinates in narrow-base and normal walking at time-points Pre (solid), Post1 (dash-dot), and Post2 (dotted). The x-axis in the Cartesian coordinates represent one gait cycle. One gait cycle in polar coordinate is [0 2π]. *Black; normal walking, red; narrow-base walking.*

FWHM

None of the FWHMs were significantly affected by training. FWHMs were found to be smaller in narrow-base compared to normal walking in the synergies associated with weight acceptance of the dominant leg and the stance phase of the non-dominant leg (synergies 2 & 3;

($F_{1,20} = 92.86$, $p < 0.001$; $F_{1,20} = 17.06$, $p < 0.001$, respectively, Figure 5.8). In contrast, FWHM of synergies associated with heel strike appeared to be greater in narrow-base compared to normal walking, but only significantly so for the non-dominant leg (synergies 4 & 5, $F_{1,20} = 2.198$, $p = 0.153$; $F_{1,20} = 8.603$, $p = 0.008$ respectively, Figure 5.8). In none of the synergies, FWHM was significantly affected by the interaction of time-point x condition ($P > 0.05$).

CoA

None of the CoAs were significantly affected by training ($p > 0.05$). CoA of synergy 1, associated with dominant leg stance, occurred significantly earlier in narrow-base compared to normal walking ($F_{1,20} = 6.005$, $p = 0.015$, Figure 5.8). CoAs of synergy 3 associated with non-dominant stance leg and synergies 4 and 5, associated with heel strike, were delayed in narrow-base compared to normal walking ($F_{1,20} = 9.832$, $p = 0.002$; $F_{1,20} = 22.109$, $p < 0.001$; $F_{1,20} = 18.308$, $p < 0.001$, respectively, Figure 5.8).

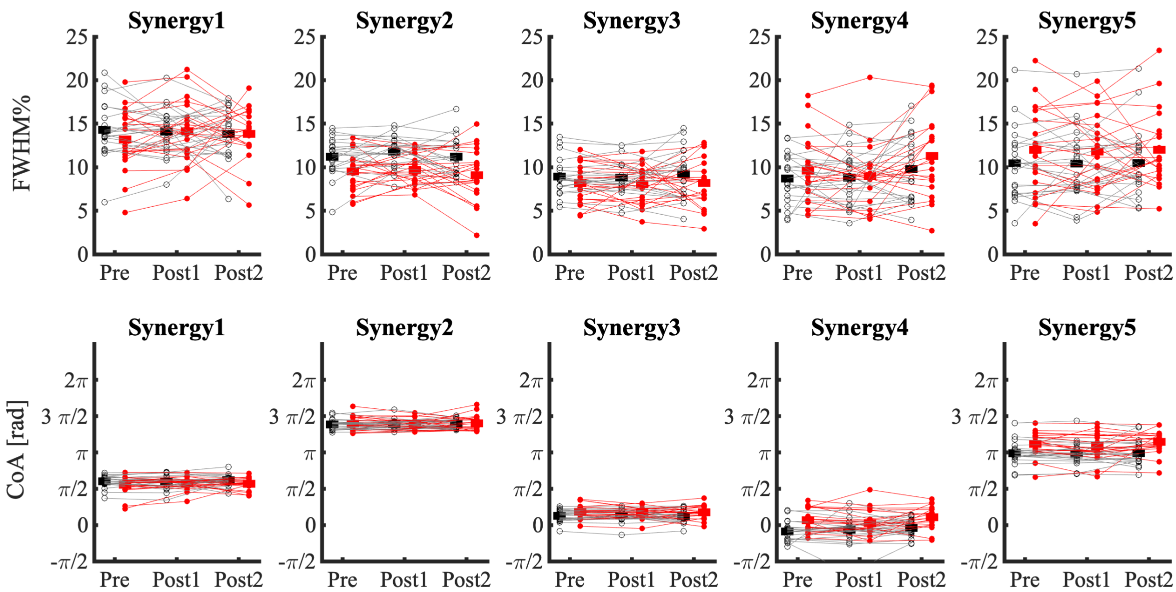

Figure 5.8: FWHM and CoA of five synergies, in narrow-base and normal walking at time-points Pre, Post, and Post2. Thin lines represent individual subject data. Thick horizontal lines indicate means over subjects. Black; normal walking, red; narrow-base walking. One gait cycle in polar coordinate is [0 2∏].

Discussion

We investigated the transfer of the effects of standing balance training to gait control, by studying gait adaptations to narrow-base walking. We previously reported improvements in robustness and performance of standing balance after short- and long-term standing balance training [206], but here we found no improvements due to training in foot placement error and CoM kinematics during normal or narrow-base walking. Participants adapted their CoM kinematics to foot placement constraints, despite not managing to step consistently within the virtual beam. These adaptations to narrow-base walking did not show an interaction with training. Furthermore, participants adapted to narrow-base walking by modifying activation profiles of their synergies. Standing balance training did not affect these activation profiles, nor their adaptation to narrow-base walking.

In line with literature [194], our participants appeared to control CoM movements more tightly during narrow-base walking than during normal walking, as reflected in a lower CoM displacement and velocity. However, again in line with literature [194], variability of CoM displacement was larger in narrow-base walking. This larger variability might reflect on-line corrections of the CoM trajectory to match it to the constrained foot placement. Confronted with a narrower base, older adults reduced mediolateral CoM displacement and velocity more than young adults [194]. This stronger response might be caused by more cautious behavior, and apparently our balance training did not alter it. Possibly, gait training has more potential to affect balance confidence in gait [220].

Five synergies described leg muscle activity across narrow-base and normal walking, together accounting for 87% of the variation in muscle activity. In spite of differences in muscles measured, participant age and walking conditions between studies, (the number of) these synergies resemble results reported in previous literature [37,43,221–224]. In our analysis, we kept the muscle weighting in these synergies, constant between conditions and time-points. Participants adapted the activation profiles of these synergies to the gait condition, but no effects of training were observed.

The FWHM of the activation profiles were different between conditions but were not affected by training. An increase of FWHM has been suggested to increase the robustness of gait [40], but in narrow-base walking our participants only increased the FWHM of the activation profile associated with non-dominant leg heel strike (synergy 5), although a similar tendency could be observed for the dominant leg (synergy 4). These adaptations of the activation profiles may reflect increased activity to enhance control over foot placement or to enhance robustness of the new stance leg in preparation for weight transfer. In contrast, participants shortened the FWHM of the activation profiles associated with the stance phase of the non-dominant leg and weight acceptance of the dominant leg. These synergies share muscle activation related to weight acceptance and the change in the activation profiles is mainly visible in a slower build-up of muscle activity (Figure 5.7). This may reflect a slower weight acceptance by the new support leg, possibly related to the lower activation peak during push-off observable in synergy 1.

The CoA of the activation profiles was different between conditions but was not affected by training. Narrowing step width led to an earlier CoA of the activation profile associated with dominant leg stance (synergy 1) and delayed CoAs of the activation profile associated with dominant and non-dominant leg heel strikes (synergies 4 and 5). Earlier CoA in the dominant leg stance phase appears to be a consequence of the reduction in activation during the second peak of the activation profile (Figure 5.7). This reduction in activation would reflect a decrease in muscle activity related to push-off and possibly reflects a more cautious gait. The earlier CoA of the activation profile associated with heel strike reflects a more sustained activation following a slower build-up (Figure 5.7). Again, this may be related to a more cautious walking but also to active control over CoM movement during the stance phase. The latter is supported by the fact that muscles that would contribute to mediolateral control, specifically tibialis anterior, peroneus longus and gluteus medius are part of these synergies. To check that changes in CoA and FWHM of the activation profiles were not due to changes in duration of gait phases, we assessed single support and double support times as percentages of the stride times and no effects of condition were found.

We studied effects of a balance training program of only 3-weeks. For transfer of acquired skills to a new task, it may be necessary that a high skill level is achieved and possibly more than 3 weeks are needed. Improved gait parameters were reported after 12 weeks of balance training [191]. Therefore, a longer duration of training might have led to changes in mediolateral gait stability.

In conclusion, older adults adapted mediolateral CoM kinematics during gait to narrow-base walking and this was associated with changes in synergies governing the activation of leg muscles. However, we found no evidence of a change in control of mediolateral gait stability, nor of these adaptations as a result of balance training.

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Chapter 6

Summary and General Discussion

Summary

Maintaining balance during daily-life activities appears easy, but a complicated and extensive active control system is established in our body to make it possible. The system becomes less accurate, slower, and less precise as we age, but the control is adaptive, which may allow us to deal with this. The extent to which the aging balance control system can cope determines the safety of movement in daily life, and depends on health conditions, level of physical activity, and environmental challenges. In chapter two, adaptations to balance challenges were compared between young and older adults. We compared the center of mass velocity in unipedal standing, as an indicator of balance performance, between age groups and between surfaces differing in compliance. We found that center of mass velocity was significantly higher in older adults than young adults and increased with increasing surface compliance in both groups. Furthermore, we found that between bipedal and unipedal standing, both young and older adults depressed their Soleus H-reflex gains. With increasing surface compliance in unipedal stance, young adults decreased H-reflex gains and increased co-contraction of their ankle muscles. In contrast, older adults did not adapt H-reflex gains or co-contraction to surface compliance. Overall, older adults showed higher co-contraction duration and lower H-reflex gains compared to young adults. This may suggest that older adults are more challenged, even when standing unipedally on a rigid surface. Hence, they show decreased H-reflex gains and increased co-contraction, but cannot further adapt them when surface compliance increases.

The findings of chapter two indicate that high co-contraction and low H-reflexes in older adults are insufficient for proper balance control on compliant surfaces. Co-contraction stiffens the ankle joint and, as such, may reduce unwanted oscillations at the ankle [118]. It serves as a feedforward strategy to compensate for delays in neural transmission and sensory-motor processing, and it may decrease electromechanical delays to allow faster corrective responses [93]. Lower H-reflex gains are thought to reflect reliance on supraspinal control associated with cortical engagement and attention to the balance task [97]. This may be needed in older adults to compensate for sensory and motor losses and may be needed in young adults when balancing on more compliant surfaces. Our results suggest that modulating H-reflex gains and co-contraction are not sufficient for proper balance control when balance is challenged. Lower H-reflex gains and higher co-contraction at more compliant surfaces in younger adults coincided with a larger center of mass velocity (poorer performance). Overall, H-reflex gains were lower and co-contraction was higher in older adults than young adults. This raises the question of

whether these two aspects of balance control (co-contraction and H-reflex gain) can be modified in older adults by balance training?

In chapter three, we reported the results of a study on balance training using unstable and compliant surfaces in older adults. Mediolateral balance robustness during unipedal stance (time to balance loss in unipedal standing on a robotic platform with decreasing rotational stiffness) improved after a single training session with no further improvement after three weeks of training. This improvement was maintained at the final measurement point two weeks after the last training session. Balance performance (mediolateral velocity of the center of mass during unipedal balancing) was also improved, in a challenging condition (perturbed stance) already after a single session and in a less challenging condition (unperturbed stance) after three weeks.

The co-contraction duration of the ankle muscles was increased after long-term training in both unipedal conditions. H-reflex gains and paired reflex depression did not significantly change with training. Cross-sectional analyses showed that lower H-reflex gains and higher co-contraction duration coincided with higher robustness and better performance at several measurement timepoints. However, changes in robustness and performance were uncorrelated with changes in co-contraction duration, H-reflex gain, or paired reflex depression after short- or long-term training. Therefore, H-reflex gain and paired reflex depression of the Soleus muscle and co-contraction duration of the ankle muscles could not be identified as determinants of improved balance. Although the increased ankle muscle co-contraction suggests its functional role in balance control, the lack of associations between changes in balance robustness and performance and changes in control at the ankle level may suggest that balance improvements may additionally be determined by changes in reactive balance control.

For this reason, in chapter four, we studied in more detail the effects of training on reactive or feedback control after perturbations of unipedal balance. We assessed the (rate of change of the) whole-body angular momentum and center of mass acceleration, and, activity of axial, proximal and distal muscles in responses to balance perturbations. Perturbations were angle-controlled surface rotations applied while subjects were in unipedal stance. After training, center of mass acceleration following the perturbations was decreased, indicating improved performance. In addition, the rate of change in angular momentum was decreased, indicating a more efficient recovery from the perturbation. However, the changes in angular momentum did not contribute to repositioning the center of mass position by a counter-rotation strategy as expected, but were used to reorient the upper body to a vertical orientation with upper and lower body aligned. Changes in the temporal activation of the muscle synergies coinciding with changes in kinematics revealed an altered temporal activation in synergies including ankle and trunk muscles suggesting that after training, participants optimized their muscle activation depending on the direction of the perturbation. This implies an improved feedback in addition to feedforward control.

Finally, in chapter five, we assessed whether improved standing balance control would transfer to gait in normal or challenging conditions. We studied adaptations in neuromuscular control of gait between normal and narrow-base walking. We found no effects of balance training or interactions between training and walking conditions on gait stability parameters or synergies. We conclude that changes in synergies and gait stability parameters to narrow-base walking may be related to a more cautious weight transfer to the new stance leg to enhance control over the center of mass movement in the stance phase.

To summarize, this thesis showed that:

- Older adults are less able to adapt their balance control to environmental demands.
- Balance control can be trained in older adults.
- Acquired balance control does not transfer to gait.

Regarding the underlying mechanisms we found that:

- Co-contraction modulation was different between age groups, it was not adapted to differences in surface compliance in older adults, but it increased with training. Co-contraction was higher in older adults with a better balance performance, but changes in co-contraction were not associated with changes in balance performance or robustness.
- H-reflex gains were lower in older than young adults, and unlike young adults, older adults did not adapt the H-reflex gain to differences in surface compliance. Although, both H-reflex and paired reflex depression were modulated between stance conditions in older adults, H-reflex and paired reflex depression did not change with training in older adults, and changes in H-reflex and paired reflex depression were not associated with changes in performance or robustness.
- Synergy activation and strategy of balance recovery were changed by training. Altered activation of ankle and upper body, and reduced rate of change of angular momentum were observed. Changes suggested improved feedback control and were not coincided with accelerating the center of mass, but with reorienting the upper body to a vertical orientation with upper and lower body aligned.

Most balance control studies compare groups with different levels of skill, health, or age in a cross-sectional design. Cross-sectional studies might give us an insight into the factors that coincide with proper balance performance. However, neural and mechanical adaptations in older adults may compensate for balance impairments. Therefore, such a cross-sectional approach can mask intrinsic balance impairments and be misleading when findings are used to guide the design of training protocols. Therefore, a longitudinal approach, as used in chapters 3-5 of this thesis may be more useful to obtain insight into mechanisms that contribute to proper balance control. For example, based on chapter 2, one might conclude that high levels of co-contraction as found in older adults are disadvantageous for balance performance. However, combined with results in chapter 3, one would conclude that this high level of co-contraction is likely adaptive and performance in older adults would probably be worse without it. Significant correlations between changes in neuromuscular control and changes in balance performance and robustness would provide support for the importance of the control aspects involved as determinants of proper balance control. However, such statistically significant correlations were not found. This may be because improvements in balance robustness and performance are multifactorial and different changes in the control may have been dominant in different participants. This would require a multivariate analysis and, consequently, a much higher number of participants than realized in the present thesis. In the following, I will discuss the findings on aspects of balance control studied in this thesis as well as reflect on aspects not studied here.

Reflexes

In chapter 2, we found lower H-reflex gains in older adults than young adults and worse balance performance in older adults. In addition, in older adults, we found no H-reflex modulation in unipedal stance on surfaces with increasing compliance, while young adults down-modulated H-reflex gains with increasing compliance. In chapter 3, we found a negative correlation between H-reflex gains and balance robustness and performance. All in all, these data suggest that lower H-reflex gains are beneficial in balancing tasks such as studied here. This is in line with previous studies on the effect of balance challenges and on learning balance tasks in young adults [72,97,109]. However, older adults seemed unable to further down-modulate their H-reflex gains in response to an increased challenge, nor did the H-reflexes changes as a result of training. This could be because a further decrease in gains is impossible to achieve or because the necessary plasticity is lacking in the short-term (adaptation to surface compliance) and long-term (training). Decreases in the H-reflex are thought to reflect a reduced effect of spinal feedback circuitry on motor control, coinciding with increased supraspinal control [56,148].

However, supraspinal mechanisms also affect the excitability of the alfa motoneuron pool and therefore the H-reflex gain. This hampers the interpretation of the H-reflex. Therefore, measurements of paired reflex depression were added in chapter 3 to provide an insight into peripherally induced inhibition which would more exclusively reflect changes in peripheral feedback. However, we also did not find any changes in paired reflex depression. Furthermore, H-reflex conditioned by Transcranial Magnetic Stimulation in young adults reduced after four weeks of balance training [56]. Reduced H-conditioned coincided with improved perturbed stance balance control, suggesting that mainly supraspinal adaptations caused improved balance control after the training. Therefore, in future studies on balance training, adding Transcranial Magnetic Stimulation to study changes in corticospinal excitability would have added value [56].

Co-contraction

In chapter 2, we found more co-contraction in older adults than young adults but worse balance performance in older adults. In addition, in older adults, we found no modulation of co-contraction in unipedal stance on surfaces with increasing compliance, while young adults increased co-contraction with increasing compliance. In chapter 3, we found an increase in co-contraction with training, which coincided with improved balance performance and robustness. However, the changes in co-contraction were not correlated with the changes in balance performance and robustness. Higher co-contraction on more compliant surfaces in younger adults suggests that higher co-contraction is a compensatory strategy used already at lower levels of challenge in older adults. So again, we might suggest that higher co-contraction in older adults is compensatory. However, older adults were unable to quickly adjust the level of co-contraction to differences in surface compliance. On the other hand, with training, co-contraction did increase, which may have contributed to the increased robustness and performance. All in all, these findings suggest that co-contraction is a compensation strategy that older adults use (and can learn to use) to improve their balance.

Muscle synergies and balance strategy

Since trunk and ankle muscle responses were altered after the training and these responses precede those of the hip muscles, results may suggest that the ankle strategy became more effective. Possibly, the increased ankle co-contraction allowed for better initial responses to the perturbations. While in the present thesis, these strategies were not studied in young adults, it has been found that young adults rely more on the ankle strategy and less on the counter-rotation strategy than older adults[180]. Furthermore, in chapter two young adults were found to increase ankle muscle co-contraction with increased challenge in unipedal balancing, while older adults did not show such a modulation. With training, older adults did increase ankle muscle co-contraction in a similar unipedal balancing task. It is still unclear if increased co-contraction was helpful or not, however the increased co-contraction resembled what young adults did. So, overall results suggest that after training the participants used a recovery strategy that might be closer to that in young adults.

Other mechanisms

Although we studied several aspects of balance control, a comprehensive analysis was not possible. We observed changes in co-contraction, muscle synergies and strategies after the long-term training. However, changes in these control aspects were not significantly correlated to changes in robustness and performance, possibly because these aspects are not the only ones affected by training.

Training effects could be co-determined by changes in supraspinal balance control mechanisms. Therefore, in future work we will analyze the EEG data collected, to investigate changes at the supraspinal level. We aim to assess whether changes in balance performance and balance strategies after training are related to changes in cortico-muscular coherence.

Aging leads to a reduction in explosive force production due to loss of fast-twitch muscle fibers [225,226] and this negatively affects balance control [227]. Balance training has been shown to improve muscle force production and specifically increase the rate of force generation [228]. This may have contributed to balance improvements found, but was not addressed in this thesis.

Balance training also was shown to result in sensory reweighting in young adults. An initial and very fast improvement in balance control with balance training was associated with up-weighting of visual information, while further improvement seemed associated with down-weighting of proprioceptive information gains [25]. The latter may be related to the decreases in H-reflex amplitudes with training that were shown previously in young adults [72,97,109], but this was not found here in older adults, neither in H-reflex gains nor in paired reflex depression. The effects of training on visual and vestibular reweighting were not addressed in this thesis.

Several studies showed a strong correlation between concern of falling and balance performance [229,230]. It has been shown that poor balance performance is mediated by changes in the allocation of attention in the presence of concern of falling [229]. The concern of falling is reduced after training in older adults, which is associated with improved balance performance [230,231]. We used the Falls Efficacy Scale International (FES-I) questionnaire at pre, post2, and retention time-points [232](pre and post1 measurements were performed on the same day). FES-I outcomes are on a scale of 16 to 64, with 16 indicating minimum concern about falling and 64 severe concern about falling.

A repeated measures ANOVA indicated that concern of falling was affected by balance training ($F_{2,42}$= 4.37, P = 0.039; Figure6.1). Post-hoc analysis showed that concern of falling was not significantly changed immediately after the training program but was decreased at retention (t = 2.16, p = 0.072; t = 2.82, p = 0.022, respectively)

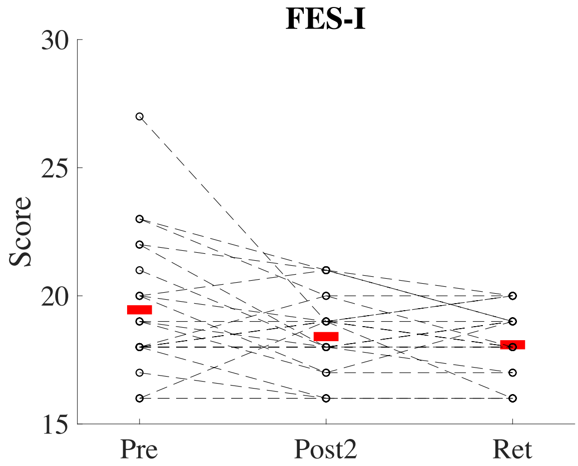

Figure 6.1: FES-I scores at different time points. Each of the lines between timepoints represents the score of a single participant.

It has previously been shown that less fear is correlated with better performance [229], and we found a decreased concern of falling in older adults between the first and the last measurement. Our findings suggest that decreased concern of falling after the balance training may explain why balance control was improved. At baseline, we found that older adults with a lower concern of falling showed higher robustness (r = 0.554, p = 0.007). However, we did not find a correlation between the change in concern of falling and the changes in robustness and performance after training.

Implication for clinical practice

The training program studied in this thesis was focused on standing balance, and our results showed the task specificity of the outcomes. Older adults with different characteristics and in a different range of ages require specific functionality. Some very old adults seek to function appropriately at their home, stand up, sit down, and get out of bed. Others with an active lifestyle might look for more control and stability in mobility to be socially engaged or even participate in sports. Given the specificity of effects, training protocols should be designed to fit the needs of the participants.

This thesis showed that balance training improves balance control in older adults in line with a large body of literature [120,233,234] and proper balance control is crucial in the daily-life activity of older adults. Without proper balance control, falls may occur and these can have a large impact. Moreover, balance training leads to a lower level of concern of falling in older adults, as addressed above. This likely improves the quality of life and may lead to higher physical activity, which in itself may protect against falls and fall-related injuries [235].

While some may argue that older adults with a higher level of activity expose themselves to the risk of falls, others may argue that staying active even when involving higher risk may be better than being inactive, isolated, but physically secured. There is no comprehensive overview of the health benefits of exercising and physical activity versus the risk of falls in older adults. However, several studies showed the benefits of having an active lifestyle [236–239], and an occasional fall that does not result in an injury might be an acceptable outcome. Notably, older adults are vulnerable to infections, with high mortality, as is clear in the current pandemic [240]. Hospitalization after injury strongly elevates the risk of infection [241]. Therefore, fracture and other major injuries associated with fall may be the more relevant outcome. Physical activity has been shown to reduce fracture risk after falls in older adults [236–238]. A supervised training program within an appropriate range of challenges might compromise between risks and benefits of physical activity in older adults. Future studies should address effects of training on neuromuscular control of balance in a range of activities and preferably also assess on fall-related injury as an outcome.

Supplementary material; chapter 2.

Supplementary material 2.1

In the experiment, arm markers moving forward were blocked from view in a few participants. Therefore, for consistency, all data on arm movement were omitted from analysis. To make sure that this has not affected our conclusions, the analysis has been redone with arms included for those subjects without missing markers for trials with peripheral nerve stimulation ($n_{old} = 8$, $n_{young} = 10$) and without peripheral nerve stimulation ($n_{old} = 7$, $n_{young} = 8$). The effect of surface compliance and age on vCOM (arms included) for the trials without/with peripheral nerve stimulation are mentioned below:

For trials without peripheral nerve stimulation (Surface Compliance, $F_{(3,39)} = 4.540$, $p = 0.008$; Age, $F_{(1,13)} = 12.206$, $p = 0.004$).

For trial with peripheral nerve stimulation (Surface Compliance, $F_{(3,48)} = 7.010$, $p < 0.001$; Age, $F_{(1,16)} = 16.758$, $p < 0.001$).

Moreover, results for trials analyzed with and without inclusion of the arms were highly correlated as shown in the figure below.

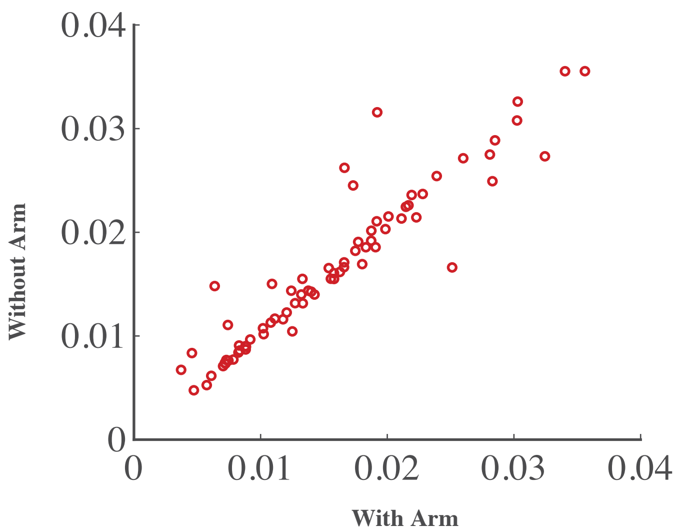

Figure S2.1. supplementary: Scatter plot of the trials without peripheral nerve stimulation, x-axis vCoM of the whole body, y-axis vCoM of the whole-body excluding arms.

Supplementary material 2.2

As an alternative explanation of down-modulation of H-reflex gains with stance condition and surface compliance, the decreased H-reflex could be due to increased bEMG. To test this explanation, we normalized the bEMG to bEMG during Bipedal standing (Figure S2.2.). The normalized bEMG did not support the alternative interpretation, as there were no significant age and stance effects, nor an interaction effect of age and stance condition on normalized bEMG ($F(1,18) = 0.408$, $p = 0.531$, $F(1,18) = 3.603$, $p = 0.074$, $F(1,18) = 0.408$, $p = 0.531$ respectively).

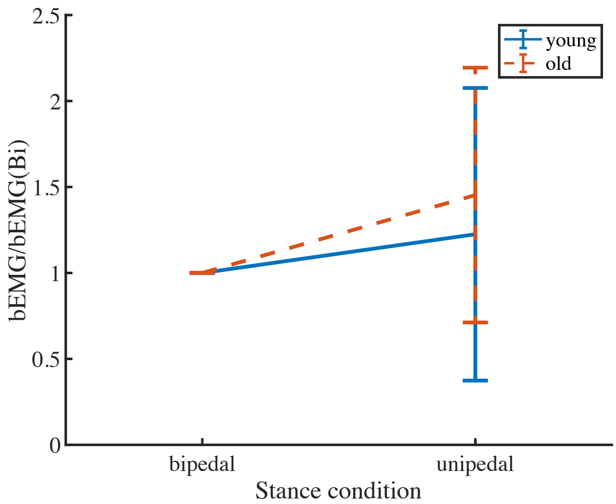

Figure S2.2. Normalized bEMG in 2 stance conditions and in young and older adults

We also repeated the analysis by normalizing bEMG to bEMG at 10% mgh, which is the condition with the highest EMG amplitudes and hence closest to the MVC. Again, no effects of age or stance condition or an interaction of age and stance condition were observed on bEMG ($F(1,17) = 0.496$, $p = 0.491$, $F(1,17) = 0.104$, $p = 0.752$, $F(1,17) = 1.243$, $p = 0.280$ respectively).

Moreover, we normalized the bEMG at all surface compliances to bEMG during Bipedal standing. The results did not support the alternative explanation of the down-modulation in young adults, as there were no age or surface effects, nor an interaction effect of age and surface compliance on normalized bEMG ($F(1,17) = 0.010$, $p = 0.921$, $F(3,51) = 2.703$, $p = 0.055$, $F(3,51) = 2.632$, $p = 0.06$ respectively; Figure S2.3.).

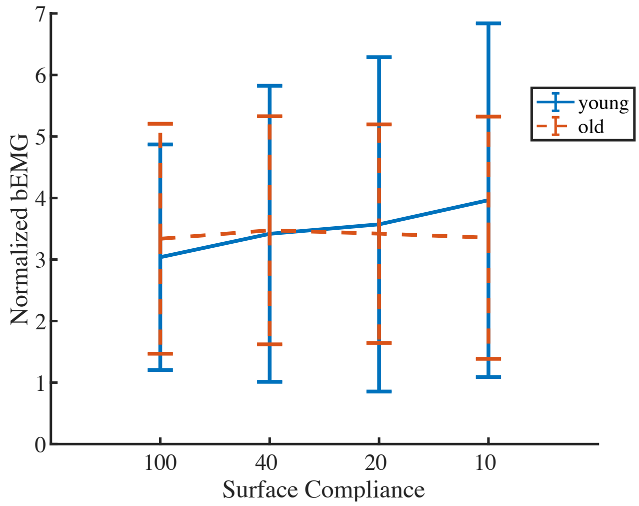

Figure S2.3. Normalized bEMG at 4 surface compliances and in young and older adults.

Supplementary material; chapter 3

Supplementary material 3.1

Progression criteria were based on the researcher's observation during the training sessions; if participants were able to perform the task for 60 seconds, the difficulty level would be increased (Figure. S3.1.). The progression plan was as follows:

Figure. S3.1. Balance training materials.

Table S3.1. Guideline for training progression

Number	Exercise	Duration/Frequency
Warm-up		
1	head rotations	rotate head to either side 5 x 3 repetitions
2	back stretching	stretch 3 x 3 repetitions
3	trunk rotations	5 rotations to both sides 3 repetitions
Exercises		
4	balancing - one leg stance (when possible) - switch the legs - unstable surfaces	3 x 60 seconds 2 repetitions
5	balancing eyes-closed - one leg stance (when possible) - switch the legs - unstable surfaces	3 x 60 seconds 2 repetitions
6	displacement of weight - one leg stance - switch the legs - unstable surfaces	3 x 60 seconds 2 repetitions
7	passing/throwing around a ball in groups of 4 fitness ball - one leg - unstable surface 2 kg ball - one leg - unstable surface alternative approaches: - make the circle bigger. - with back towards each other in order to induce more trunk rotations.	5 rounds both directions 3 repetitions
8	pass the big ball around while stopping it on foot and role it to the other person. fitness ball - one leg - unstable surface 2 kg ball - one leg - unstable surface	5 rounds both directions 3 repetitions

Supplementary material 3.2

For reflex measurement it is generally recommended to elicit H-reflex between 15-40% of M_{max} [123,124] , while we elicited H-reflex at H_{max}, in line with our previous study. However, for 20 out of 22 participants H_{max} was at less than %40 of M_{max} (see Figure S3.2. supplementary).

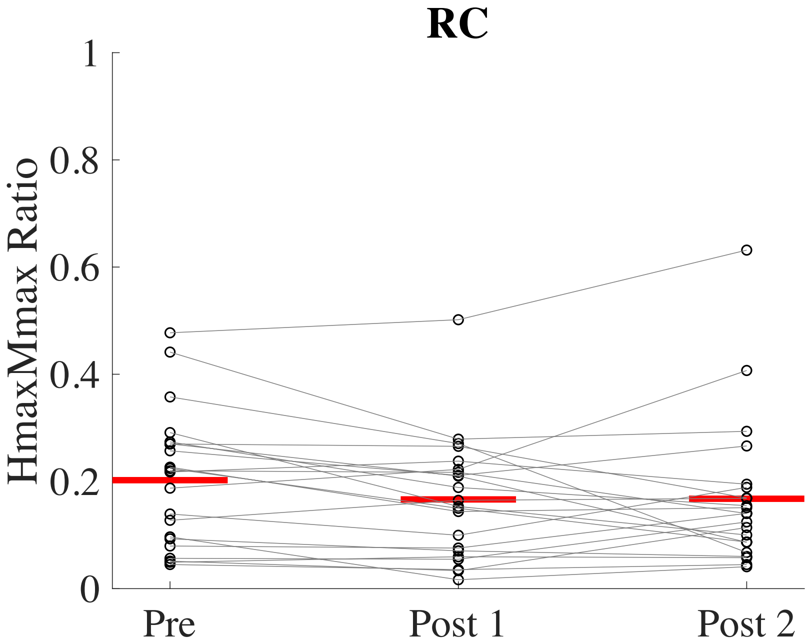

Figure S3.2. supplementary: H_{max}/M_{max} ratio obtained from the recruitment curve (RC) for all participants at 3 time points, indicates H_{max} less than %40 of the M_{max} for 20 out of 22 subjects.

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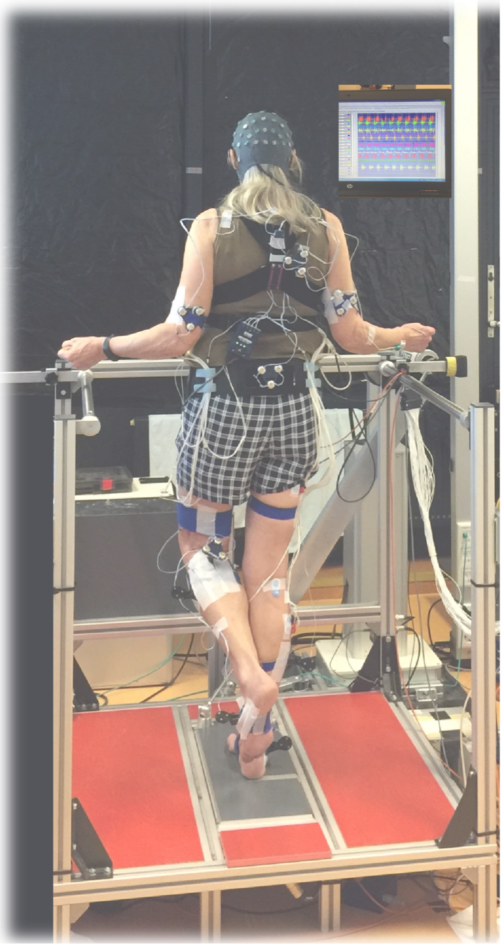
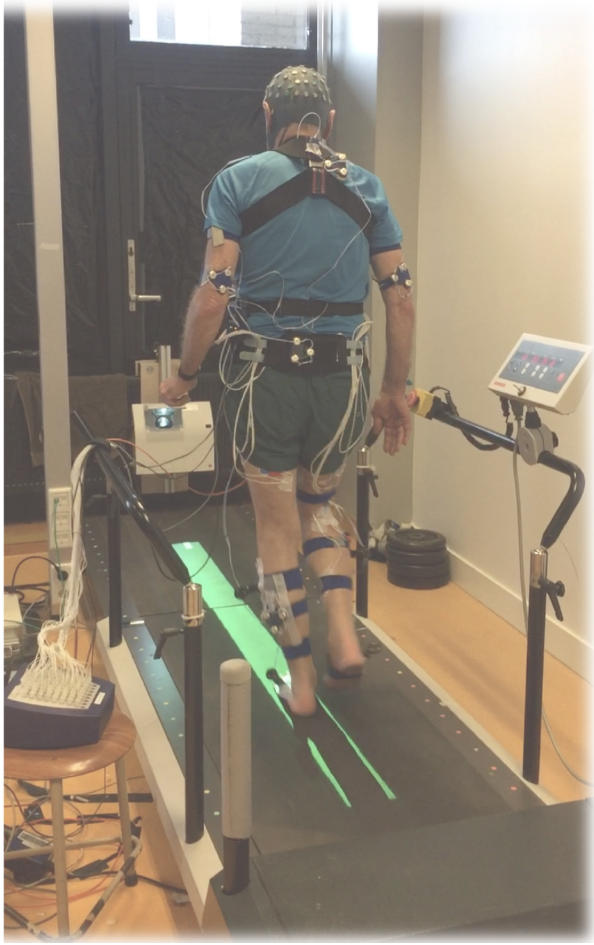

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