Advanced age slows learning to balance with imposed sensorimotor delays independent of vision

By Christian van der Zalm, 445251

Supervisor: Patrick Forbes

Abstract word count: 249

Character count with spaces of thesis: 49.017

Abstract

To maintain standing balance, the brain must overcome inherent time delays between self-generated motor commands and sensory feedback. Through natural aging, these delays increase due to gradual changes in muscle force generation and neural processing. Consequently, humans must adapt their balance control to maintain upright bipedal posture during advanced age. In this study we characterized the effect of aging on sensorimotor learning during standing balance with imposed delays. We further explored the reliance on vision for balance learning given the reported reliance that older adults place on vision during standing balance compared to young adults. We used a robotic balance simulator to impose a 250 ms delay between balancing motor actions and whole-body sway in healthy older (> 65 years, N = 20) and young (<30 years, N = 20) participants. Older and young participants were randomly divided into two groups that trained to balance with the delay with vision or no vision (N=40). Our results show that compared to young participants, older adults learned to balance with the delay at a slower rate and with greater instability. Importantly, regardless of the visual condition that was trained, both old and young participants could generalize and transfer this learning to the untrained visual conditions to improve stabilization. These outcomes suggest that although aging can decrease the rate and effectiveness of learning to overcome delays during balance, older adults are able to regain stability even with eyes closed. This has promising implications of improving balance with delay training for fall-prone adults.

Introduction

As the average age of the global population increases, age related problems will play a bigger role in society (Khan, 2018). One of the greatest problems of aging is the risk of falling (Sharif et.al., 2018). The consequences of falling at an older age include higher hospitalization rates, longer rehabilitation periods and higher chance of morbidity (Khan, 2018). Moreover, even if an older adult survives a fall, most often independent mobility will never return to the level prior to the fall (Khan, 2018). The risks of falling induce a "fear of falling", also in older adults that have not yet experienced a fall themselves, which makes them more cautious and less dependent (Schoene et.al., 2018). This ultimately has one of the biggest influences on the quality of life of older adults (Schoene et.al., 2018). Therefore, it is important to investigate the underlying causes of the decreased balance stability that comes with advanced age, beginning with standing balance itself.

Bipedal upright posture is inherently unstable, as anti-gravity muscles in the torso and legs need to continuously correct our upright posture (Horak, 2006; Forbes et.al., 2018; Promsri et.al., 2020). This process is centrally controlled by the brain, which receives information about the state of the body in the environment in the form of somatosensory, vestibular and visual information (Forbes et.al., 2018). In a dynamic environment it is important to weight and integrate these modalities accordingly to their estimated reliability for the control of posture, before the brain can send accurate motor commands to the corresponding muscles (Eikema et.al., 2013). The speed by which these motor commands can be generated is limited by factors such as sensory processing and nerve conductivity, which creates a time delay between processing and motor action (Forbes et.al., 2018). This means that by nature to maintain standing balance, the brain must overcome and adapt to these inherent time delays between self-generated motor commands and sensory feedback. It has been repeatedly observed that with advanced age, sensorimotor delays increase while sensorimotor adaptation decreases (van der Kruk et.al., 2021). In addition, it has been shown that sensory receptors become less sensitive in older adults, such as higher perception thresholds for vibration of the skin. This reduces the reliability of incoming information to the brain about changes of body orientation. Moreover, impairments of the motor systems related to aging include decreased lower body muscle strength and more frequent coactivation (Wiesmeier et.al., 2015). The effect of age on postural control adaptation is often assessed using external perturbations to destabilize the body (Brauer et.al., 2001; Doumas and Krampe et.al., 2010). However, external perturbations do not represent the changes that are caused by aging, such as increased sensorimotor delays.

An alternative approach, explored in healthy young adults (Rasman et. al., 2021), is to expose participants to unexpected sensorimotor delays between the ankle-generated torque and whole-body sway. Upon first exposure to the delay participants were very unstable, but through prolonged training (~30 min) they could regain stable balance. Using the same paradigm, here we aim to investigate the effect of age on sensorimotor learning with imposed delays during balance. Decreased sensorimotor learning in older adults is frequently linked to an overreliance on visual field dependence during postural control adaptation, like perturbation adaptation (Yeh et.al., 2014). Studies that investigated the influence of removing vision during postural control have shown that older adults are able to rely more on proprioceptive and vestibular information, although these two modalities are not able to fully compensate for the absence of vision (Wiesmeier et.al., 2015). This is often explained by a reduced ability of sensory reweighting. Sensory reweighting is the process of prioritizing some sensory modalities over others, depending on the sensory relevance given the task (Assländer et. al., 2014). However, this is still under debate. Most studies that are in favour of the reduced reweighting theory were all based on altering the visual field to make the visual information less reliable for maintaining balance (Allison et. al., 2006; Yeh et.al., 2014), instead of removing visual cues by closing the eyes. These studies used external perturbations as a form of adaptation. Using imposed sensorimotor delays instead gives perhaps a better insight into to the internal sensorimotor changes caused by natural advanced age.

The aim of our study was to characterize the effect of aging on sensorimotor learning during balance and to determine the underlying mechanisms for differences in observed adaptation. We performed experiments using a robotic balance simulator where young and older participants were trained to balance in AP (anterior-posterior) direction with an imposed delay of 250 ms. In both age groups, we tested separate sub-groups of participants who trained balancing with the eyes open (vision) and closed (no vision). We hypothesized that this learning to regain balance with imposed delays would be slower in individuals of advanced age compared to younger individuals and that this learning would be less effective, given the age-related decline in sensory and motor function during standing balance. We also further predicted that the absence of vision during training would slow down learning in both young and older adults, but that this effect would be more pronounced in older individuals. This is due to current studies that have showed that older adults are more dependent on vision during balance. Moreover, we explored how participants adapted to the sensorimotor delays by looking at the frequency content of the ankle-generated torque when learning to balance with imposed delays. Overall, our findings demonstrate that older adults do show slower and less effective sensorimotor learning, but with less effect of vision than expected.

Lastly, to further investigate if learning to deal with unexpected sensorimotor delays involves adaptive motor learning, we have done experiments with two cerebellar stroke patients. The cerebellum is known as that of a predictive system, which when impaired influences the patient's ability to correctively predict their own movement and adapt to novel sensory feedback (Zimmet et. al., 2020). These studies then found that cerebellar patients compensate for this deficit by relying more on intact sensory feedback pathways, such as vision. We have conducted two experiments on cerebellar stroke patients, where they trained to balance with a fixed imposed delay of 200 ms with eyes open. We hypothesised that the cerebellar patients would have a hard time reducing their stability during training. However, we also hypothesised that the cerebellar stroke patients still would be able to regain stability for a small amount, as they would be able to compensate their predictive deficits with visual cues. Preliminary results seemed promising as the patients were able to reduce their stability, although less than healthy older adults that trained with a bigger delay. The aim of this preliminary study was to give an insight about characterizing sensorimotor learning in cerebellar stroke patients.

Materials & Methods

Population

A total of 20 healthy young participants ranging between 18 and 30 years old (23.55 ± 1.98) and 20 older adult participants ranging from 65 and 79 years old (70.1 ± 3.96) participated in this study. The exclusion criteria for both groups included: history of neurological or psychiatric disorders, acute or chronic psychoactive drug use, alcohol abuse, pregnancy, neuropathy in the lower extremities, diabetes, history of balance deficits, neuromuscular injuries, skeletal deformities or arthritis in the lower extremities. All participants gave written consent prior to the experiment and the consent form was ethically approved by The Erasmus Medical Center Medical Ethics Review Committee (MEC-2022-0239). All participants had normal or corrected to normal vision. To limit the influence of peak force generation and muscular atrophy, older adults with a BMI above 35 were excluded from the study. Table 1 shows the information of all training groups. For the power calculation we used an alpha = 0.05 and an effect size = 0.4, based on the variance and effect sizes of whole-body sway learning (pre vs. post) behaviour and the interaction with different delayed conditions (Rasman et al., 2021). As we have a sample size of 10 participants for each training group, the calculated power of this study is 0.6.

Table 1 Training group characteristics

Training Group	Age (mean ± SD)	Sex ratio (Male/Female)	BMI (mean ± SD)
Older No Vision	69.5 ± 4.30	5/5	25.65 ± 2.94
Older Vision	70.8± 3.58	5/5	25.34 ± 3.27
Young No Vision	23.7 ± 2.54	4/6	22.68 ± 3.54
Young Vision	23.2 ± 1.34	4/6	21.69 ± 6.14

The two cerebellar stroke patients that were part of the preliminary study were recruited and selected by a neurologist. Both patients were scored following the International Cooperative Ataxia Rating Scale (ICARS), which is a standardised clinical rating system for ataxia treatment trials, to make sure the patients were able to perform the experiment (Storey et. al., 2004). Both cerebellar stroke patients finished their initial rehabilitation period prior to the experiment. One cerebellar stoke patient was tested 3 months after the stroke happened in the left part of the cerebellum (59-year-old male, ICARS: 2) and the other cerebellar stroke patient was tested 12 months after the stroke happened at the right part of the cerebellum (69-year-old female, ICARS: 0). Both patients gave consent prior to the experiment.

Experimental setup

Participants stood on a custom-designed robotic balance simulator that simulated the normal mechanics of a standing person as an inverted pendulum pivoting around the ankle joints (see Figure 1). Previous studies have shown that this model approximates the relationship between the ankle-generated torques and the resulting body position (Luu et al., 2011, Rasman et. al. 2021, Forbes et al. 2016). For real-time implementation, the simulator used a continuous transfer function (see Equation 2) that was converted to a discrete-time equivalent using the zero-order hold method.

$$I\ddot{\theta} - m_m gL\theta = T \qquad \textit{(Equation 1)}$$

$$\frac{\theta}{T} = \frac{1}{Is^2 - 0.971mgL} \qquad \textit{(Equation 2)}$$

Here, θ represents the angle of the centre of mass relative to the ankle joint from vertical, where plantarflexion of the ankle is defined as a positive angle and dorsiflexion as a negative angle. T is the ankle-generated torque, I is mass moment of inertia of the body, g is gravitational acceleration, L is the distance between the centre of mass and ankle joint, m is the total mass of the participant and m_m is the effective mass of the participant, calculated as $0.971m$. This effective mass was used by assuming that the inverted pendulum pivots around the ankle, whereby the mass of the feet is removed.

The simulation ran on a real-time system (PXI-8880; National Instruments, TX, USA) at 500 HZ and controlled a 2 kW servo motor (ECMA-J11020S4, Delta, Taiwan; maximum continuous torque: ~6170 Nm; angular resolution of ~0.0000054°) that was attached to a rigid backboard through a linear actuator (Y-H1116165P09152A; Rollon, Italy). The AP motion of the backboard was software limited to 6 degrees forward and 3 degrees backwards. Outside these limits, the program gradually increased the simulated stiffness to the point that participants couldn't rotate any farther in that direction independent of the ankle-generated torques they produced, therefore providing passive support of the body at that angle. Active torques applied in the opposite direction allowed the participant to leave the limit and resume balance (Rasman et. al., 2018). The maximum velocity of the backboard was limited to 12 rads/s. The software used to control the system was custom designed in LabVIEW (2019a). Medium dense foam cushions were attached to the harnesses of the backboard for participant comfort that could be height-adjusted to the waist and shoulders. The participant was secured to the harnesses using seatbelts and an extra pair of medium dense foam cushions on the front. The anterior-posterior positions of the shoulder and hip harnesses were adjusted to hold the participant in an upright position and minimize the AP momentums and shear forces at rest. To measure the ankle-generated forces, participants stood bare foot on a support surface with a built-in force plate (AMTI BP400×600; Watertown, MA, USA), which was fixed horizontally throughout the experiment. The participant controlled their own whole-body orientation by generating plantarflexion and dorsiflexion torques around the ankle.

During the delay conditions, a delay of 250 ms was imposed between the ankle-generated torques of the participant and the resulting body position, taking into account the internal 4 ms delay of the system. The position signal of the backboard motor and the torques generated by the participants were recorded with a data acquisition board (PXI-6289; National Instruments, Austin, TX, USA) at a sampling frequency of 500 Hz. During the experiment, participants wore noise cancelling headphones (WH-1000XM3 Noise Cancelling Headphones, Sony, Japan) to reduce auditory cues produced by the motors and other external sounds. Nature sounds were played on the headphones to minimize sounds arising from the motors and communication with the participant was implemented via the headphones. During no vision conditions (see Experimental protocol), participants wore a blindfold (black, EAN 8719538123090), while the lights of the room were turned off.

Before starting an experiment, we measured the mass, height, hip joint width and centre of mass height of each participant. Hip joint width was calculated as the medio-lateral distance between the greater trochanters multiplied by 0.532 (Bennett et. al., 2016). Hip joint width was used to define the location of the feet and were marked on the force plate with tape at the beginning of the experiment for reproducibility over the different trials and conditions. The centre of mass was measured by laying the participant on their back on a plank with their ankles on the border of the plank. The plank was then rolled over a stick until the participant and the plank were in equilibrium. The distance between the ankles at the border and the location of the stick was taken as an approximation of the centre of mass of the participant.

Experimental protocol

The experiment consisted of 5 phases: (1) a familiarisation phase, (2) a baseline phase, (3) a pre-condition phase, (4) a training phase, and (5) a post-condition phase. During the familiarisation phase, participants were encouraged to experience the range of motion and the maximum velocity of the system. They were also instructed to explore the relationship between the plantar/dorsiflexion of the ankle and the accompanying change in body position. During the experiment, the participant was instructed to move out of the limits as quickly as possible when balance was lost. Following the familiarization phase, participants performed baseline balancing trials where they were instructed to maintain a comfortable upright position for 1 minute, while having their eyes open (vision condition) and while wearing a blindfold (no vision condition). Participants then performed pre-condition trials where a 250 ms delay was imposed between the ankle-torques and the body position. To minimize any learning behaviour before training, we limited these trials to 30 seconds and performed them twice for both vision conditions (vision and no vision) in alternating order. Subsequently, participants were randomly allocated in one of the two training groups, vision training or no vision training, while controlling for similar distribution of sex within each group. The training phase consisted of eight 5 minute trials, with a break every two trials. More breaks were taken if requested by the participant. The post-condition phase followed the same structure as the pre-condition phase. The starting condition for the baseline, pre-condition and post condition phases were all randomly selected per participant, using a random number generator in Matlab 2021b.

Experimental protocol cerebellar stroke patients

The experiments conducted on the two cerebellar patients followed the same base paradigm, but the delay was reduced to 200 ms and the training time was reduced to seven 4 minute trials. Moreover, the software limits of the robotic balance simulator were reduced to 4 degrees forward and 2 degrees backward. Both cerebellar stroke patients trained with vision.

Data analysis

All data analysis was performed in Matlab 2021b. To quantify how balance behaviour improved before, during and after training, we estimated the variance of whole-body sway velocity throughout each trial by extracting the velocity signal when the participant was balancing inside the software limits, with a resolution of 2 second windows. As long as the participant maintained balance inside the limits during this time window, that velocity signal was used for the analysis. This approach was also used to examine the longest time the participant was able to maintain balanced per minute and the variance of the ankle-generated torque.

To compare the differences in balance learning between the different training groups, we fitted exponential learning curves to the whole-body sway velocity variance, the longest time the participant balanced within limits and the variance of the ankle-generated torque. Fitting to the training data was done by using an exponential function (a*exp(-x/b)+c), from which the time constant (b) was extracted. The effect of training was defined as the value of the fit at the last minute of training. This approach was chosen over taking the last point of the non-fitted training data, as the fitted function has an averaging effect and minimizes the effect of variability at the end of training. To examine the relative effect of training between training groups, we normalized the learning curves by dividing the data by the maximum value of the whole training period.

To further examine the actual motor behaviour of the participants during training, we applied a Fourier analysis on the ankle-generated torque during training per minute. For the Fourier analysis, we used the built-in fast Fourier transform function of Matlab 2021b. The torque signal was extracted as long as the participant was able to balance inside the limits for at least 5 second time windows, similar to the time window extraction method that was used for the whole-body sway velocity variance. Five second windows were chosen to increase the resolution of the Fourier signal. Per minute, the mean was taken over all 5 second windows that the participants were inside the limits.

To investigate the changes in relative frequency content in the ankle-generated torque during training, we approximated the transition frequency changes in the normalized frequency spectrum. The normalization of the power was done by dividing the frequency signal by the maximal power per minute. The transitions frequency, the frequency at which the power rapidly decreases, was determined by fitting the low and high frequency content separately, based on the methods used in a study by Carriot et. al. (2014). For the low frequency content, the mean was taken of the normalized frequency spectrum from 0.2-0.6 Hz. For the high frequency content, the slope of the normalized power between 0.6-10 Hz was fitted following the form of a n-order low pass equation: 1/sqrt((x/w0)^(2*n)), with w0 being the resonance frequency. We then approximated the transition frequency by the intersection of these two fits.

Statistical analysis

For the statistical analysis we used IBM SPSS Statistics 28. To examine the effect of the imposed delay on the whole-body sway velocity variance, we applied a mixed design ANOVA with repeated measures on the baselines and pre-conditions for both vision and no vision trials, with the imposed delay as the within participant variable. To test our hypotheses about the effects of age, vision and any interaction effects on balance learning, we applied a two-way ANOVA on the time constants and the effect of learning of the sway velocity variance, the longest period time within limits and the torque variance for the different training groups. Moreover, to examine the effect of training for all vision conditions, a mixed design ANOVA with repeated measures was used to compare the pre-training and post-training mean velocity variance per vision condition, with the training time as the within participant variable. To evaluate the presence of transfer between training in one vision condition to the untrained vision condition, we applied mixed design ANOVA with repeated measures on the post-training mean sway velocity variances, with the vision condition as the within participant variable. Furthermore, to to investigate the frequency content change in the ankle-generated torques, we applied a two-way ANOVA on the transition frequencies and the mean power at the start and end of training for the different training groups.

Results

Twenty older and twenty young adult participants were instructed to stand quietly on a robotic balance simulator and maintain balance for the duration of all trials, while experiencing imposed delays between the torques generated around their ankles and subsequent whole-body sway in AP direction. Before the trails with the imposed delay, all participants performed baseline trials where there was no imposed delay, to represent normal balance (See Figure 1). All participants performed one minute baseline trials in both no vision and vision conditions. During the baseline trials, all participants maintained stable balance with limited sway velocity variance. For both age groups, there was a significant main effect of vision ($F(1, 38)= 8.454$, $p=0.006$), with sway velocity increasing by 194% on average in no vision conditions, but there was no significant main effect of age ($F(1, 38)= 0.047$, $p=0.830$).

After baseline conditions, we examined how novel imposed delays increase postural oscillations without any prior training by exposing all participants to a fixed delay of 250 ms for four 30 s trials, alternating between no vision and vision conditions. During these pre-learning trials, participants balanced with large sway velocity variance, for both vision (older: 12.52 ± 3.323 $(°/s)^2$), young: 10.23 ± 5.994 $(°/s)^2$) and no vision conditions (older: 10.73 ± 3.006 $(°/s)^2$, young: 9.662 ± 3.691 $(°/s)^2$). We compared the sway velocity variance between baselines and pre-conditions for each vision condition using repeated measures mixed design ANOVA (see methods). The effect of imposing the delay on the balance control as compared to baseline conditions was significant for both vision ($F(1,38)= 218.476$, $p<0.001$) and no vision conditions ($F(1, 38)= 355.278$ $p<0.001$) for all age groups. There was again no main effect of age.

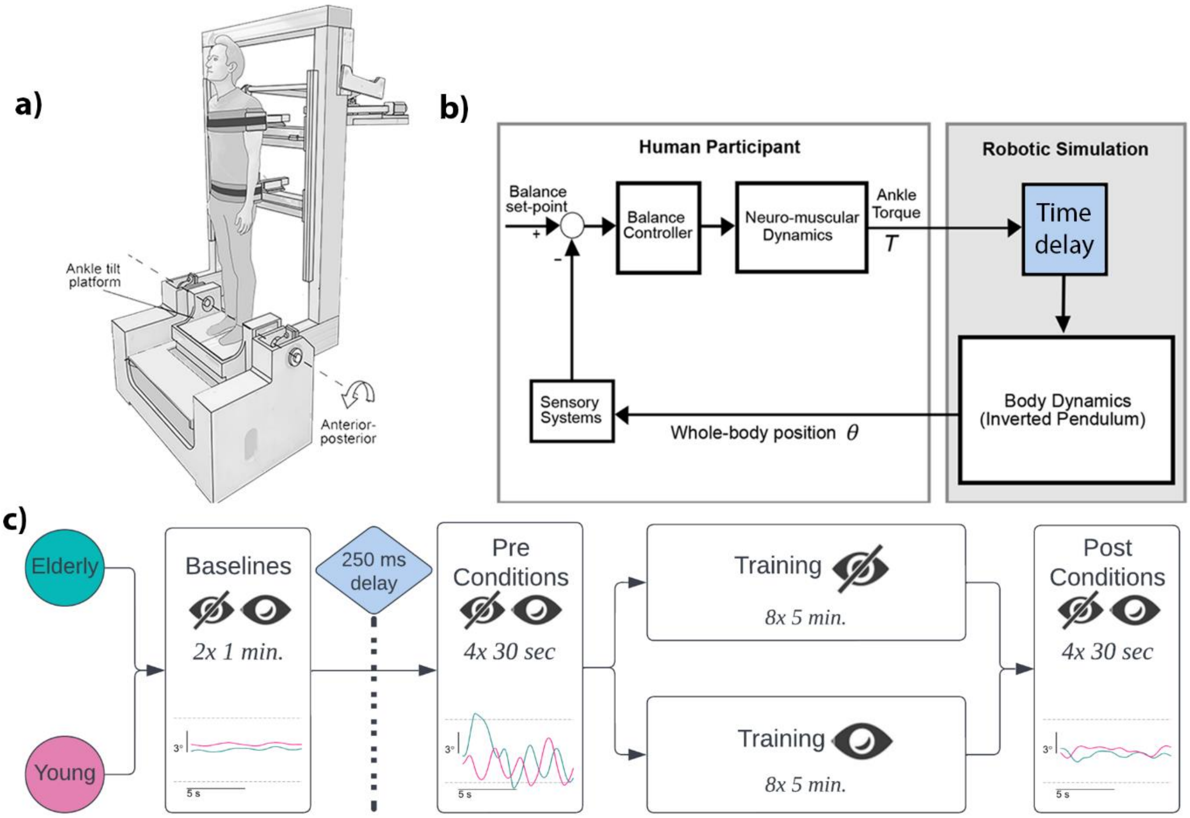

Figure 1 Experimental setup a) *Schematic of the balancing robot. The motors moved the body of the participant in AP direction depending on the ankle-generated torques the participant applies to the force plate (see methods).* **b)** *Schematic of the robotic balance simulation control loop. The delay was imposed between the ankle torques generated by the participant and the movements of whole-body via the motors.* **c)** *Schematic of the experimental protocol. Older and young adults performed the same set of baselines, pre and post conditions, both with vision and no vision. Both age groups were divided in a training group that trained with an imposed delay with eyes open (vision) or eyes closed (no vision). The traces represent the whole-body sway traces of a representative older (green) and young (pink) participant over all stages of the experiment.*

After the pre-condition trials, both older and young participants were randomly allocated into groups that trained to balance with the imposed delay with vision (N =10) or no vision (N = 10). Throughout training, almost all participants reduced their sway velocity variance and increased the longest time of balance within the limits. Exponential fits to the velocity variance (all R^2>=0.94) capture the differences across each group (see Figure 2a inset): At the beginning of training, older adults seem to have on average a higher sway velocity variance than younger adults. For the younger adults there seemed to be no difference in learning behaviour between the participants that trained with or without vision. In contrast, it seemed that on average older adults that trained without vision showed less learning than the older adults that trained with vision. We quantified these groups by extracting the time constant of learning (τ) and the sway velocity variance at the end of training (the effect of training). There was a significant effect of age on the extracted time constants of the fitted learning curves of the velocity variance (older: 15.25 ± 10.88 minutes, young: 9.219 ± 7.269 minutes) ($F(1,36)= 4.229$, $p=0.047$), but no main effect of training vision condition (vision trained: 10.29 ± 7.226 minutes, no vision trained: 14.16 ± 11.41 minutes). This suggests that older adults have a slower learning rate when dealing with the imposed sensorimotor delay, independent of the presence of vision. Similarly, older adults also showed a significant higher sway velocity variance at the end of training (2.775 ± 2.709 $(°/s)^2$) independent of training vision condition, which means that they had a reduced effect of training compared to young adults (0.7711 ± 0.5008 $(°/s)^2$) ($F(1,36)=10.495$, $p=0.003$). By normalizing the sway velocity variances, we investigated how the relative effect of training differed across groups. The asymptotic behaviour of the learning curves in Figure 2a, seems to show that there were again 3 distinct groups, but distributed differently; now it seems that older adults that trained with vision converged to the same velocity variance as that of the young participants that trained without vision. Indeed, these normalized results showed a significant effect of age on the relative effect of training ($F(1,36)= 64.100$ $p=0.018$) as older adults still had an increased normalized sway velocity variance at the end of training (0.2463 ± 0.2013) compared to young adults (0.1297 ± 0.0734). The time constants of the normalized sway velocity variance stayed the same. This means that not only are older adults slower at learning to deal with the imposed delay, also the effect of learning is reduced.

To further quantify the balance learning behaviour of the participants, we looked into the longest time per minute that the participants were able to maintain balance without falling into the limit of the robotic balance simulator (Figure 2b). Similar to the normalized velocity variance, it seemed that on average older adults that trained with vision converged to the same effect of learning at the end of training as that of younger adults that trained without vision. At the end of training older adults had significantly less time they could maximally maintain balance within one minute without falling into the limits (47.72 ± 12.31 s), compared to the younger adults (5.63 ± 5.972 s) ($F(1,36)=7.258$ $p=0.011$). There was also a main effect of training vision condition, as participants that trained without vision had a reduced time of maximally maintaining balance within one minute without falling into the limits (48.42 ± 11.83 s), compared to participants that trained with vision (54.93 ± 7.606 s) ($F(1,36)=4.916$ $p=0.033$). There was no interaction effect of training vision condition with age. Moreover, older adults had higher time constants (14.30 ± 10.20 minutes) of the fitted learning curves to the longest time within limits (all $R^2>=0.89$), compared to younger adults (7.013 ± 7.773 minutes) ($F(1,36)=6.405$, $p=0.016$). Lastly, to get an insight on the motor behaviour of the learning process, we looked at the normalized torque variance (Figure 2c). The relative effect of training from the normalized learning curves (all $R^2>=0.91$) was significant for age ($F(1,36)=82.241$, $p=0.037$), but not significant for training vision condition ($F(1,36)=3.813$ $p=0.059$), although it seemed that on average the effect of training was reduced for participant that trained without vision.

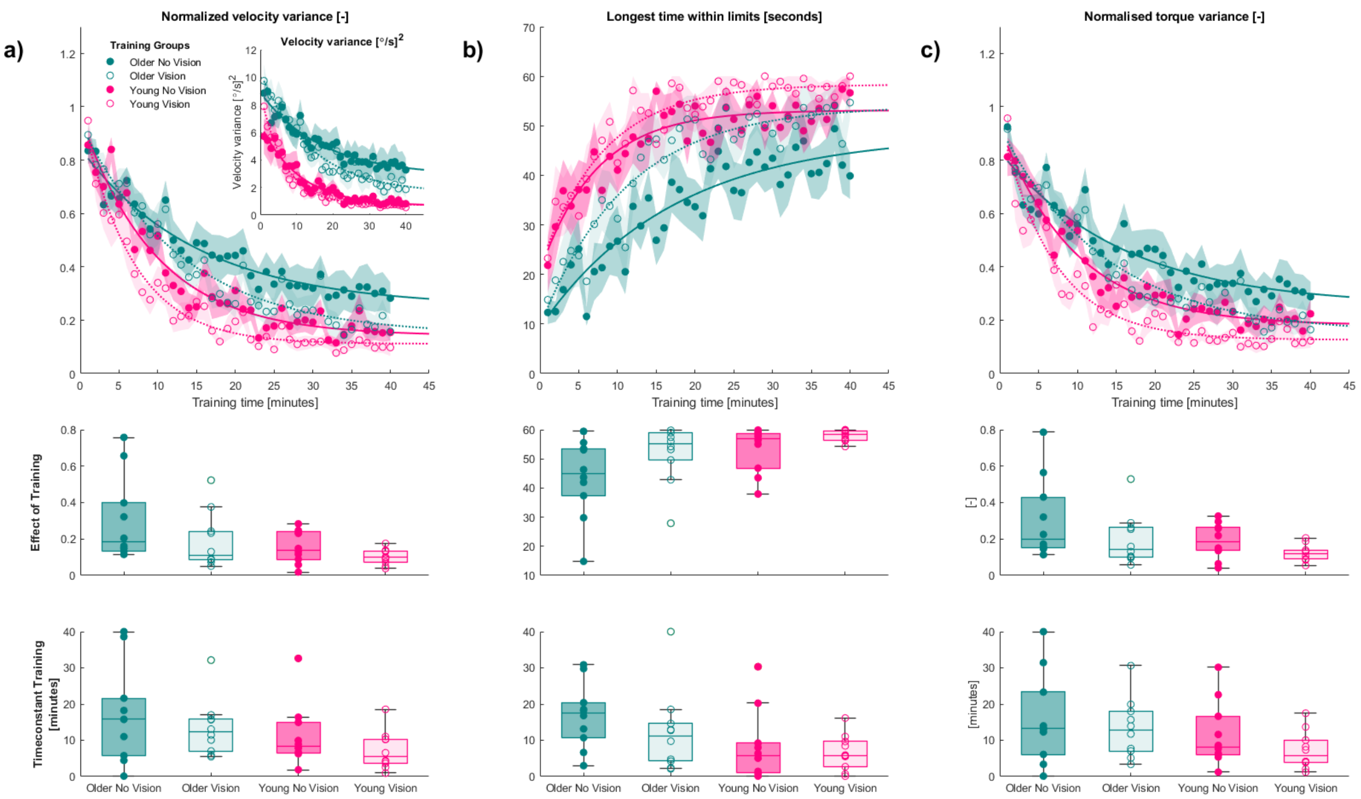

Figure 2 Training with an imposed delay a) *The whole-body sway velocity variance for all training groups during training per minute and the fitted exponential learning curves. The lower box plots show the extracted effect of training and time constants for each training group. The time constants were limited to 40 minutes. The normalized sway velocity variance graph is unitless as the data is normalized to the maximum sway velocity variance during training. The coloured areas show the standard error of the mean (SEM) per minute of each group.* ***b)*** *The longest time the participants balanced within a minute during training before falling into the limits of the robotic balance simulator for all training groups and the fitted exponential learning curves. The lower box plots show the extracted effect of training and time constants for each training group. The time constants were limited to 40 minutes The coloured areas show the standard error of the mean (SEM) per minute of each group.* ***c)*** *The variance of the normalized ankle-generated torque during training for all training groups per minute and the fitted exponential learning curves. The lower box plots show the extracted effect of training at the end of training and time constants for each training group. The time constants were limited to 40 minutes. The normalized sway torque variance is unitless as the data is normalized to the maximum torque variance during training. The coloured areas show the standard error of the mean (SEM) per minute of each group.*

After training with the imposed delay in their respective vision condition (vision or no vision), each participant performed four 30 s post-training trials with the same fixed delay, alternating between vision and no vision conditions. This allowed us to investigate if the learning effects on balance from training in one vision condition transfers to the other vision condition. Figure 3 shows the sway velocity variance comparisons between the pre- and post-training trials for all groups. For all training groups, the sway velocity variance post training was significantly reduced compared to pre training in both vision and no vision conditions (all $p<0.001$), without interactions of age or training condition. This indicates that training with an imposed delay improves balance behaviour, independent of the vision condition trained in. This speaks to the generalizability of training with an imposed delay. Comparing the two vision conditions during the post-training trials within participants, there was a significant interaction with training condition ($F(1,36)=10.519$, $p=0.003$). This means, although training in one condition improves stability in both vision conditions, balance behaviour is better in the conditions that is trained in.

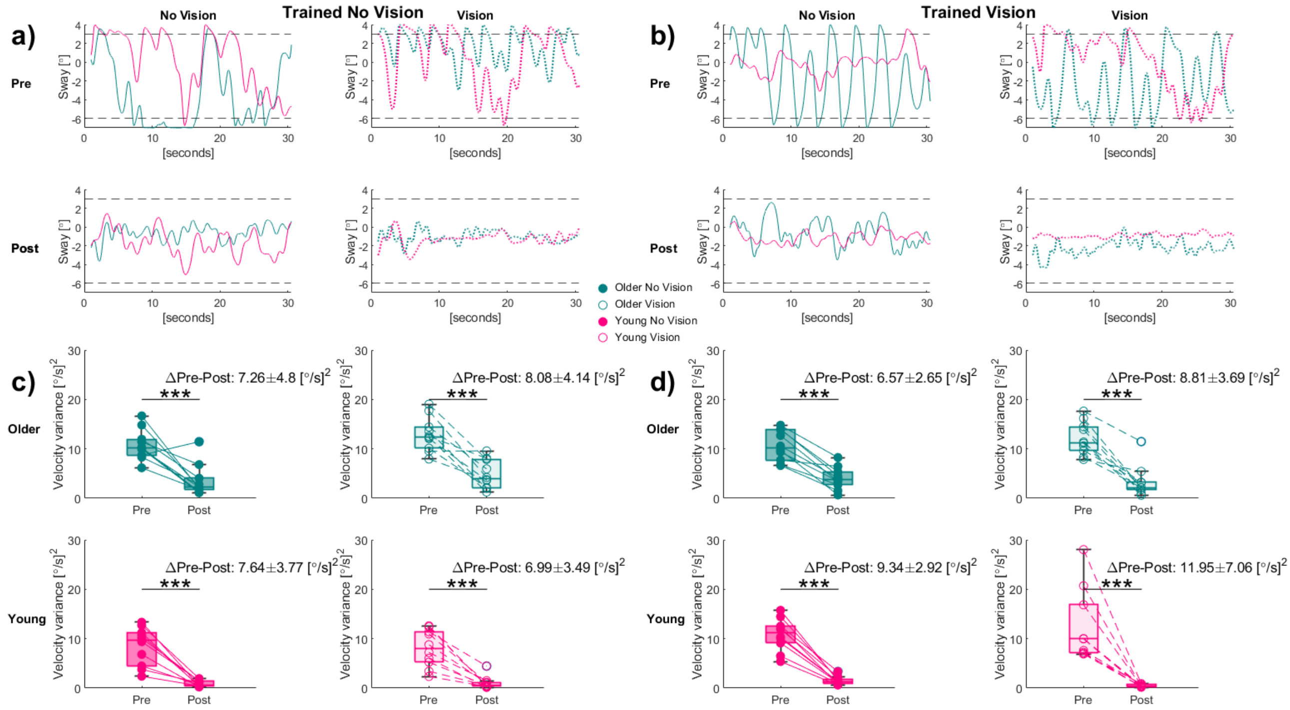

Figure 3 pre and post training comparisons of sway velocity variance a) *Sway traces of representative older (green) and young (pink) adult participants that trained without vision, in the pre- and post-training phases of the experiment for both vision conditions (vision: dashed line, no vision: filled line).* ***b)*** *Sway traces of representative older (green) and young (pink) adult participants that trained with vision, in the pre- and post-training phases of the experiment for both vision conditions (vision: dashed line, no vision: filled line).* ***c)*** *Boxplots of the sway velocity variances of the training groups that trained without vision, pre and post training for both vision conditions. In the older adults no vision pre-post comparison, it shows that one participant did have a higher sway velocity variance after training compared to the pre-trials. We decided to include this participant as they showed learning during the training period.* ***d)*** *Boxplots of the sway velocity variances of the training groups that trained with vision, pre and post training for both vision conditions.*

To further investigate the change in motor behaviour that accompanied the strategy that participants used to learn to balance with an imposed delay, we examined the frequency spectrum of the ankle-generated torque. We reasoned that if adaptation to sensorimotor delays would involve integrating the delay into the internal model of the motor system, then a decrease in the higher frequency content of ankle-generated torque would be observed. We compared the mean power of the time-frequency spectrum of the torque at the beginning and end of training. During the first 5 minutes of training there was no significant effect of age or training condition on the mean power of the torque. Both young and older participants reduced their gain over the course of training, although this happened quicker in young participants. In the last 5 minutes of training young participants had a significant lower gain compared to older adults (older: 2108 ± 301.99 $(Nm)^2$, young: 591.4 ± 301.99 $(Nm)^2$) ($F(1,36)=12.618$, $p=0.001$). This effect was independent of training vision condition. Normalizing the frequency spectrum of the torque to the maximum power per minute gives an indication how the relative contributions of different frequencies changes across the frequency spectrum. By fitting both the high frequency content and the low frequency content, we calculated the transition frequencies at which the power starts to decrease rapidly (see Methods). Older adults showed a higher transition frequency, both at the start (older: 0.651 ± 0.026 Hz, young: 0.510 ± 0.026 Hz) ($F(1,36)=14.705$, $p<0.001$) and end of training (older: 0.701 ± 0.022, young: 0.555 ± 0.022 Hz)($F(1,36)=22.765$, $p<0.001$), independent of training vision condition. The transition frequency change between the start and end of training was significant for all training groups ($F(1,36)=6.029$, $p=0.019$), without interactions of age or training condition.

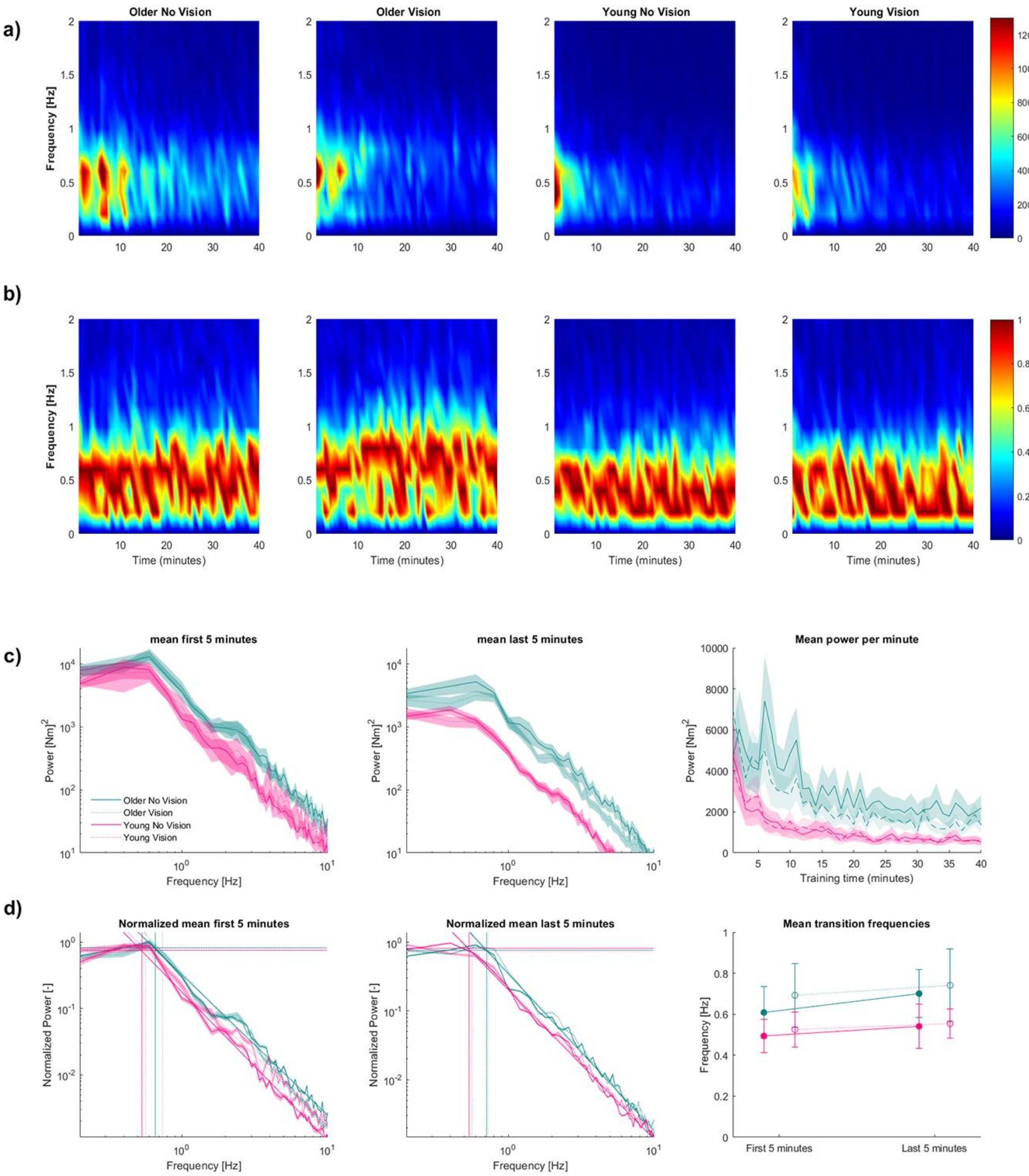

Figure 4 Fourier analysis of the ankle-generated torque a) *Time-frequency power spectrum of the ankle-generated torque over the training period for all training groups up to 2 Hz.* ***b)*** *Time-frequency power spectrum of the ankle-generated torque over the training period for all training groups up to 2 Hz, normalized to the maximum power per minute.* ***c)*** *The power spectrum of the first and last 5 minutes of training. The most right panel shows the mean power over all frequencies upto 2 Hz for the whole training period. The coloured areas are the standard error of the mean (SEM).* ***d)*** *The mean power spectrum of the first and last 5 minutes of training normalized to the maximum power per minute. The horizontal lines represent the mean power upto 0.6 Hz and the diagonal lines represent the decreasing slope of the higher frequency power from 0.6 Hz to 10 Hz, as done by Carriot. et al. (2014). The vertical lines represent the transition frequencies, where these two fits intersect. The most right panel shows the mean transition frequencies for all training groups with the errorbars representing the standard error of the mean (SEM).*

Discussion

The aim of our study was to characterize the effect of aging on sensorimotor learning during balance and to determine the underlying mechanisms for differences in observed adaptation. We expected that older participants would be slower and less effective at learning to balance with imposed delays, given the age-related decline in sensory and motor function during standing balance. Moreover, we expected that the absence of vision during training would have a bigger effect on older adults, as previous studies suggest that older adults are more dependent on vision during balance. Our results indicate that older adults seem to learn to balance with unexpected sensorimotor delays at a slower rate and less effective than young individuals. However, the absence of vision during this learning process had less of an effect on the older adults than expected.

Older adults learn slower and less effective than young adults, but vision has less of an effect than expected

From our outcomes it seems that while older age can decrease the rate and effectiveness of learning to overcome delays in the control of balance, the relative contribution of visual signals to this learning does not change with age. As all training groups, independent of age or training condition, were able to transfer their learning both with eyes open as closed, that means that that this learned behaviour is potentially generalizable. This suggests that older adults can sufficiently use and switch to other sensory modalities, such as vestibular and somatosensory feedback channels to compensate for the lack of vision. Our finding seems to indicate that older adults are able to use sensory reweighting during balance, which is the process of prioritizing some sensory modalities over others, depending on the sensory relevance given the task (Assländer et. al., 2014). This finding contrasts against the outcomes of previous studies that showed that older adults are less flexible in sensory reweighting and are over reliant on visual field dependence during postural control adaptation (Yeh et.al., 2014). However, the over reliance on visual cues in older adults is still under debate. For example, a study by Wiesmeier et.al. (2015) perturbed young and older adults by pseudo-random tilts of a support surface the participants stood on. Based on feedback model simulations, they actually showed that older adults even prefer proprioceptive information over visual cues during balance. Another possible reason behind the minor effect of visual deprivation in the older adults during balance is that with age the acuity of all sensory channels goes down, including visual acuity (van der Kruk et. al., 2021). The noise on these visual feedback channels, would make visual cues less reliable for maintaining postural control.

The question remains in what way the body achieves stability when dealing with sensorimotor delays. We propose that there are two possible ways the brain can adapt to these delays. One of them is integrating the delay into the motor control. This would result in a change in frequency content of the motor commands, especially higher frequencies would relatively contribute less to the motor control. However, our results indicated that all participants increased the frequency content of their ankle-generated torque, implied by the increased transition frequency where the power decreases rapidly. Moreover, we found that older adults had an overall higher transition frequency compared to younger adults, which could be due to the higher coactivation and higher ankle stiffness that is linked to advanced age (Wiesmeier et.al, 2015). These results are in line with a study, where they showed that ankle-stiffness is linked to unexpected perturbations (Le Mouel et. al., 2019). They then demonstrated that older adults show an increased ankle-stiffness even during normal balance, which provides a partial compensation for neural delays (Le Mouel et. al., 2019). Following a model-based approach, they found that and increased ankle-stiffness only provides compensation during balance if the sensorimotor gain is decreased.

This can be linked to another way to deal with sensorimotor delays, which is not integrating the delay in the motor control system, but simply reducing the feedback control gains. This would result in not a reduction of the frequency content in the motor commands, but only a reduction of the average magnitude of applied torque. Our results indeed show that most participants reduced the overall power of their ankle-generated torques during training with the delay. This outcome suggests that during learning to deal with sensorimotor delays, the delay is not integrated into the motor control system, but the participants simply learn to apply less torque to regain stability.

Further research into adaptive motor learning with cerebellar patients

To further investigate if learning to deal with unexpected sensorimotor delays involves adaptive motor learning, future studies may look at the learning behaviour in pathological populations that show compromised adaptive motor learning, such as cerebellar patients. Recent research suggests the role of the cerebellum as that of a predictive system, which when impaired influences the patient's ability to correctively predict their own movement and adapt to novel sensory feedback (Zimmet et. al., 2020). These studies then found that cerebellar patients compensate for this deficit by relying more on intact sensory feedback pathways, such as vision. Using the same paradigm on this clinical group will give more insight about the neurological mechanisms underlying balancing with sensorimotor delays. Preliminary results from two cerebellar stroke patients already suggests that cerebellar patients are able learn to balance with an imposed delays of 200 ms, although there is a large variability that can be linked to the time passed since the stroke happened (Figure 5) and the location of the stroke. It is already promising that patients are able to increase their stability with imposed delays at all. Moreover, it seems that the cerebellar stroke patients reduce their gain less than healthy participants. More data will further investigate sensorimotor learning in this clinical group and the effect of visual deprivation will be interesting to characterize, given their reliance on sensory feedback pathways.

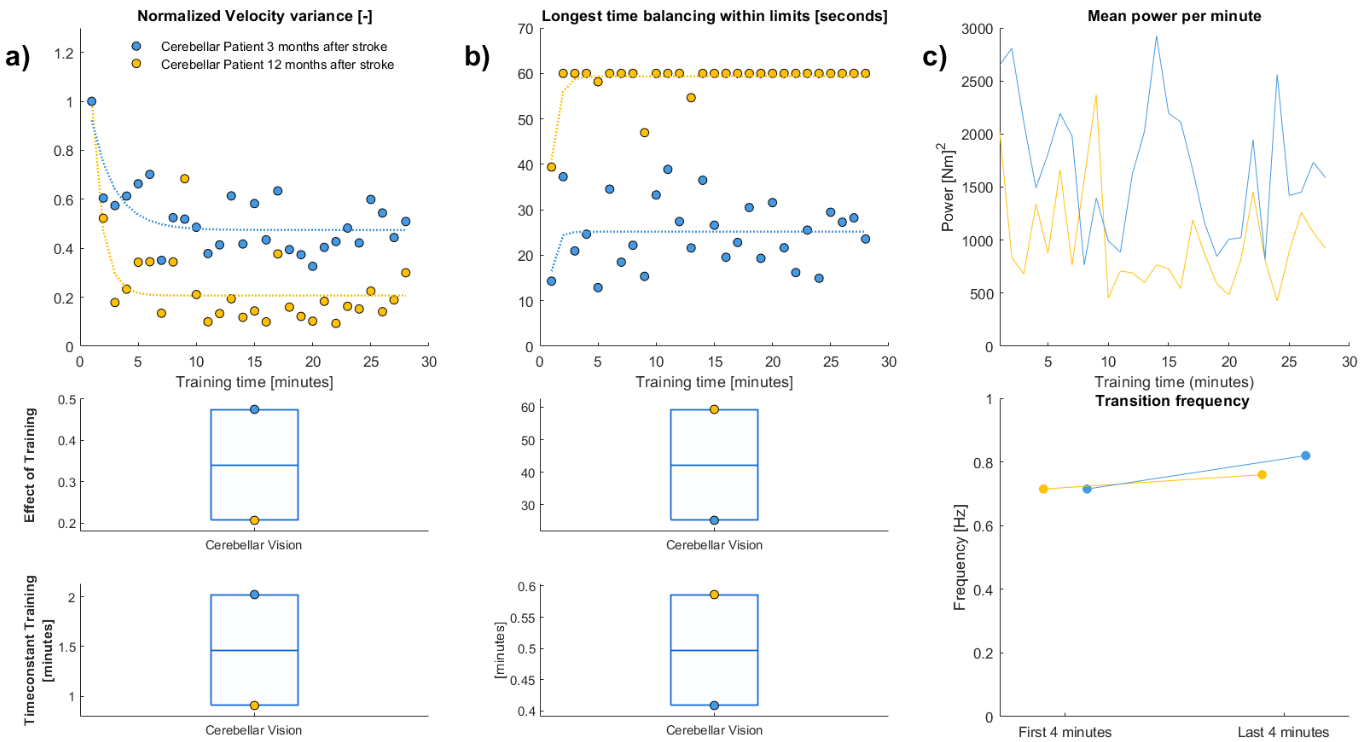

Figure 5 Preliminary results from delay training of cerebellar stroke patients a) *The normalized whole-body sway velocity variance of cerebellar stroke patients (N=2) that trained with a delay of 200 ms during 28 minutes with eyes open and the fitted learning curves. Lower panels show the relative effect of training and the time constants of the two cerebellar stroke patients.* ***b)*** *The longest time the cerebellar stroke patients were able to maintain balance before falling into the limits of the robotic balance simulator and the fitted learning curves. Lower panels show the effect of training and the time constants of the two cerebellar stroke patients.* ***c)*** *The mean power of the time-frequency sepctrum of the ankle-generated torques over all frequencies upto 2 Hz for the whole training period and the extracted transition frequencies from the normalized power spectrum.*

Limitations of current study

Some participants, mostly younger adults, did not have a hard time balancing with the imposed delay prior to training. This can be partially linked to the small delay size used in this study. As this is the first study investigating the balance behaviour of older adults with unexpected sensorimotor delays, we chose to use a smaller delay to make sure all participants would able to be reasonably stable at the end of training. We took the length of training into account given that older adults show earlier forms of muscle fatigue (van der Kruk et. al., 2021). The study from Rasman et. al. (2021) has shown that young individuals are able to learn to balance with sensorimotor delays up to 400 ms. Future experiments with higher delays than 250 ms may potentially find a larger effect of age than seen in this current study as the balance learning behaviour of the training groups may diverge.

Moreover, in the current protocol participants were able to keep balancing after they had fallen into the limits of the robotic balance simulator by pushing themselves back into the range of balancing. This created periods of balancing, mostly in the beginning of training, where some participants were slowly falling instead of actively balancing, creating artificial periods of balancing. Furthermore, some participants that fell into the limits of the robotic balance simulator applied unnecessary excessive torques trying to get themselves back to an upright position, which caused them to move back and forth at high velocities. We tried to solve this problem by limiting the data that is used for the balancing analysis to periods of a minimum of two seconds that the participant is balancing without passing the software limits. However, it was still possible for some participants that to fall from one limit to the other in more than two seconds without actively trying to balance, by for example applying small ankle-generated torques. An alternative approach would be to stop the control by the participant over the robotic balance simulator, the moment they fall into one of the limits and set them back to an upright position. However, this would first of all increase the duration of training, which would induce more muscle fatigue and potentially the effect of training. Secondly, the benefit of freely exploring the novel correlation between ankle-generated torques and whole-body sway without continuously resetting the experiment when the participant falls has potentially a greater effect on the rate of learning.

Potential advantages of Robotic balance simulation training

The outcome that older adults were able to balance with imposed unexpected sensorimotor delay, gives promising insight into the use of a robotic balance simulator as a safe form of balance training or even rehabilitation. When participants are balancing inside the robotic balance simulator, they are free to explore novel strategy that can help them to achieve stability without the risk of falling. This risk of falling makes older adults more cautious and potentially reduces exploration and adaptive abilities, as the consequences of falling worsen with age. This exploration-risk trade-off is potentially one of the many reasons why some older adults are more prone to falling. Training in a robotic simulation would give older adults a safe environment to try strategies that can help their overall balance behaviour. Currently, fall-prone older adults are given fall training to minimize the chance of fractures caused by falling, instead of trainings that would prevent them from falling (Groen et. al., 2009). There have been studies that look into preventive fall training, but these outcomes are still preliminary (Thomas et. al., 2019). Future studies may explore if there is any beneficial effect of long-term robotic balance training with imposed sensorimotor delays that transfers to off-robot balance improvement in fall-prone older adults.

Acknowledgements

I would like to thank Patrick Forbes and Brandon Rasman for the supervision and the long discussions we had. Moreover, I want to thank the Sensorimotor Neuroscience and Biorobotics Lab for all the support. Lastly, I want to thank Jacqueline van Nieuwenhuijzen for helping with recruiting most of my older participants.

References

Allison, L.K., Kiemel, T. & Jeka, J.J. Multisensory reweighting of vision and touch is intact in healthy and fall-prone older adults. *Exp Brain Res* **175**, 342–352 (2006). https://doi.org/10.1007/s00221-006-0559-7

Assländer, L., Peterka, R. J. (2014). Sensory reweighting dynamics in human postural control. Journal of Neurophysiology, 111(9), 1852–1864. https://doi.org/10.1152/jn.00669.2013

Bennett, H. J., Shen, G., Weinhandl, J. T., Zhang, S. (2016). Validation of the greater trochanter method with radiographic measurements of frontal plane hip joint centers and knee mechanical axis angles and two other HIP joint center methods. Journal of Biomechanics, 49(13), 3047–3051. https://doi.org/10.1016/j.jbiomech.2016.06.013

Brauer, S. G., Woollacott, M., Shumway-Cook, A. (2001). The interacting effects of cognitive demand and recovery of postural stability in balance-impaired elderly persons. The Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 56(8). https://doi.org/10.1093/gerona/56.8.m489

Carriot, J., Jamali, M., Chacron, M. J., Cullen, K. E. (2014). Statistics of the vestibular input experienced during natural self-motion: Implications for neural processing. Journal of Neuroscience, 34(24), 8347–8357. https://doi.org/10.1523/jneurosci.0692-14.2014

Doumas, M., Krampe, R. Th. (2010). Adaptation and reintegration of proprioceptive information in young and older adults' postural control. Journal of Neurophysiology, 104(4), 1969–1977. https://doi.org/10.1152/jn.00345.2010

Eikema, D. J. A., Hatzitaki, V., Konstantakos, V., Papaxanthis, C. (2013). Elderly adults delay proprioceptive reweighting during the anticipation of collision avoidance when standing. Neuroscience, 234, 22–30. https://doi.org/10.1016/j.neuroscience.2012.12.053

Forbes, P. A., Chen, A., Blouin, J.-S. (2018). Sensorimotor control of standing balance. Handbook of Clinical Neurology, 61–83. https://doi.org/10.1016/b978-0-444-63916-5.00004-5

Groen, B. E., Smulders, E., de Kam, D., Duysens, J., Weerdesteyn, V. (2009). Martial Arts fall training to prevent hip fractures in the elderly. Osteoporosis International, 21(2), 215–221. https://doi.org/10.1007/s00198-009-0934-x

Horak, F. B. (2006). Postural orientation and equilibrium: What do we need to know about neural control of balance to prevent falls? Age and Ageing, 35(suppl_2), ii7–ii11. https://doi.org/10.1093/ageing/afl077

Khan, H. T. A. (2018). Population ageing in a globalized world: Risks and dilemmas? Journal of Evaluation in Clinical Practice, 25(5), 754–760. https://doi.org/10.1111/jep.13071 van der Kruk E. et.al., 2021

Le Mouel, C., & Brette, R. (2019). Anticipatory coadaptation of ankle stiffness and sensorimotor gain for standing balance. PLOS Computational Biology, 15(11). https://doi.org/10.1371/journal.pcbi.1007463

Luu, B. L., Huryn, T. P., Van der Loos, H. F., Croft, E. A., Blouin, J.-S. (2011). Validation of a robotic balance system for investigations in the control of human standing balance. IEEE Transactions on Neural Systems and Rehabilitation Engineering, 19(4), 382–390. https://doi.org/10.1109/tnsre.2011.2140332

Promsri, A., Haid, T., Federolf, P. (2020). Complexity, composition, and control of bipedal balancing movements as the postural control system adapts to unstable support surfaces or altered feet positions. Neuroscience, 430, 113–124. https://doi.org/10.1016/j.neuroscience.2020.01.031

Rasman, B.G., Forbes, P.A., Tisserand R. and Blouin J-S (2018). Sensorimotor Manipulations of the Balance Control Loop–Beyond Imposed External Perturbations. Front. Neurol. 9:899. doi: 10.3389/fneur.2018.00899

Rasman, B. G., Forbes, P. A., Peters, R. M., Ortiz, O., Franks, I., Inglis, J. T., Chua, R., Blouin, J.-S. (2021). Learning to stand with unexpected sensorimotor delays. eLife, 10. https://doi.org/10.7554/elife.65085

Schoene, D., Heller, C., Aung, Y. N., Sieber, C. C., Kemmler, W., Freiberger, E. (2019). A systematic review on the influence of fear of falling on quality of life in older people: Is there a role for falls? Clinical Interventions in Aging, Volume 14, 701–719. https://doi.org/10.2147/cia.s197857

Sharif, S. I., Al-Harbi, A. B., Al-Shihabi, A. M., Al-Daour, D. S., Sharif, R. S. (2018). Falls in the elderly: Assessment of prevalence and risk factors. Pharmacy Practice, 16(3), 1206. https://doi.org/10.18549/pharmpract.2018.03.1206

Storey, E., Tuck, K., Hester, R., Hughes, A., & Churchyard, A. (2004). Inter-rater reliability of the International Cooperative Ataxia Rating Scale (ICARS). Movement Disorders, 19(2), 190–192. https://doi.org/10.1002/mds.10657

Thomas, E., Battaglia, G., Patti, A., Brusa, J., Leonardi, V., Palma, A., & Bellafiore, M. (2019). Physical activity programs for balance and fall prevention in elderly. Medicine, 98(27). https://doi.org/10.1097/md.0000000000016218

van der Kruk, E., Silverman, A. K., Reilly, P., Bull, A. M. J. (2021). Compensation due to age-related decline in sit-to-stand and sit-to-walk. Journal of Biomechanics, 122, 110411. https://doi.org/10.1016/j.jbiomech.2021.110411

Wiesmeier, I. K., Dalin, D., Maurer, C. (2015). Elderly use proprioception rather than visual and vestibular cues for Postural Motor Control. Frontiers in Aging Neuroscience, 7. https://doi.org/10.3389/fnagi.2015.00097

Yeh, T. T., Cluff, T., Balasubramaniam, R. (2014). Visual reliance for balance control in older adults persists when visual information is disrupted by artificial feedback delays. PLoS ONE, 9(3). https://doi.org/10.1371/journal.pone.0091554

Zimmet, A. M., Cao, D., Bastian, A. J., Cowan, N. J. (2020). Cerebellar patients have intact feedback control that can be leveraged to improve reaching. eLife, 9. https://doi.org/10.7554/elife.53246