Adaptation of multi-joint coordination during standing balance in healthy young and healthy old individuals


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ABSTRACT

Standing balance requires multi-joint coordination between the ankles and hips. We investigated how humans adapt their multi-joint coordination to adjust to various conditions and whether the adaptation differed between healthy young participants and healthy elderly. Balance was disturbed by push/pull rods, applying two continuous and independent force disturbances at the level of the hip and between the shoulder blades. In addition, external force fields were applied, represented by an external stiffness at the hip, either stabilizing or destabilizing the participants’ balance. Multivariate closed-loop system identification techniques were used to describe the neuromuscular control mechanisms by quantifying the corrective joint torques as a response to body sway, represented by Frequency Response Functions (FRFs). Model fits on the FRFs resulted in an estimation of time delays, intrinsic stiffness, reflexive stiffness, and reflexive damping of both the ankle and hip joint. The elderly generated similar corrective joint torques but had reduced body sway compared to the young participants, corresponded to the increased FRF magnitude with age. When a stabilizing or destabilizing external force field was applied at the hip, both young and elderly participants adapted their multi-joint coordination by lowering or respectively increasing their neuromuscular control actions around the ankles, expressed in a change of FRF magnitude. However, the elderly adapted less compared to the young participants. Model fits on the FRFs showed that elderly had higher intrinsic and reflexive stiffness of the ankle, together with higher time delays of the hip. Furthermore, the elderly adapted their reflexive stiffness around the ankle joint less, compared to young participants. These results imply that elderly were stiffer and were less able to adapt to external force fields.

Keywords: standing balance control, multi-joint coordination, adaptation, healthy elderly
Ageing is associated with impaired balance and falls (Rubenstein, 2006; Muir et al., 2010; Pasma et al., 2014b). To maintain balance, several underlying systems work together. One of the mechanisms to maintain balance is to alter the coordination of postural responses, i.e. multi-joint coordination, which is often explored by analyzing ankle strategies and hip strategies (Horak and Nashner, 1986), describing the movement in these joints. Coordination of movements around the ankle and hip joints depends on the amount of external disturbances – such as gravity and pushes having impact on the body – and on the support surface conditions (Horak and Nashner, 1986; Creath et al., 2005; Fujisawa et al., 2005). Adaptability of multi-joint coordination is an essential feature of standing balance control in order to adjust to various conditions.

Previous studies showed that the elderly tend to have altered multi-joint coordination to maintain standing balance compared to young individuals. The elderly exhibited higher cross-correlation between the upper and lower body in quiet stance, indicating that the displacements of the two body segments were less independent (Accornero et al., 1997; Gariépy et al., 2008). In addition, the elderly had a less-flexible joint coordination pattern to compensate for externally applied balance disturbances (Hsu et al., 2013). This possibly indicates that the elderly used less hip strategy and behaved more rigidly.

Various underlying mechanisms have been put forth as contributing factors of altered multi-joint coordination in the elderly. Some studies indicated that the elderly increase their reflexive stiffness around the ankle joint when they are exposed to altered sensory information (Amiridis et al., 2003; Benjuya et al., 2004), while other studies found altered intrinsic properties of muscles and tendons having impact on stiffness in the elderly (Kearney et al., 1997; Ishida et al., 2008; Cenciarini, 2010).

Detecting the underlying mechanisms of altered multi-joint coordination in standing balance is complex, as there exists substantial redundancy at the joint, muscle, and neural levels (Hsu et al.,
Multiple sensory systems contribute to balance control, i.e. the proprioceptive, visual and vestibular system. The sensory signals are integrated and processed by the central nervous system and are then used to generate corrective joint torques by precise muscle-activation patterns. Due to this redundancy, cause and effect remain unclear; increased stiffness can be due to altered intrinsic muscle properties or increased reflex activity, or it might be a compensation strategy and a result of the fact that the elderly change their dynamic behaviour to maintain stability. Multivariate closed-loop system identification techniques (CLSIT) (Boonstra et al., 2013; Engelhart et al., 2014) are required to unravel cause and effects. Applying multiple and specifically designed external disturbances on the human body allows the study of multi-joint coordination in standing balance. Multi-joint coordination is expressed in the generation of corrective joint torques as a response to body movement; i.e. the dynamic behaviour of the neuromuscular controller.

With system identification techniques, we investigated the underlying mechanisms of multi-joint coordination in standing balance in healthy young and healthy elderly. Furthermore, we investigated the adaptation to externally applied force fields. Studies in the upper and lower extremities showed that postural responses adapted when external force fields were applied (Shadmehr and Mussa-Ivaldi, 1994; Burdet et al., 2001; Franklin et al., 2003; van Asseldonk et al., 2009). In standing balance studies, it was shown that the elderly possess the ability to adapt to external disturbances, although to a lesser degree than the young (Pavol et al., 2002; Ooteghem et al., 2009). Therefore, we hypothesize that a stabilizing force field will downscale the postural responses and that this effect is more pronounced in the young than the elderly, indicating the elderly are less adaptive. In addition to existing studies, identification of the underlying factors that change multi-joint coordination with age, could provide insight into mechanisms influencing the risk of falling in the elderly (Engelhart et al., 2014). Adequate treatment of balance disorders requires unraveling the underlying primary causes and applied (adaptive) strategies.
2. METHODS

2.1 PARTICIPANTS
Fifteen healthy young (age range 20-30 years) and 14 healthy elderly individuals (age range 70-79 years) participated in the study (Table 1). Participants were excluded when they: 1) were in a dependent living situation; 2) were unable to walk a distance of 250 m; 3) presented co-morbidity (dementia, neurologic disorders, metabolic diseases, rheumatic diseases; heart failure; severe chronic obstructive pulmonary disease; 4) used medication with an influence on balance control (immunosuppressive drugs, insulin, anticoagulation); 5) were immobilized for one week during the last three months or 6) had orthopedic surgery during the last two years with unresolved pain or functional limitation.

To illustrate the state of health of the study population, several outcome parameters were measured. Cognition was assessed using the Mini-Mental State Examination (MMSE) (Folstein et al., 1975) and participants with a score lower than 26 points were not included. Physical functioning was assessed by handgrip strength and the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994). Walking speed was obtained from the four meter walking test of the SPPB. The total amount of medication was obtained by questioning the participants.

The study was performed according to the principles of the Declaration of Helsinki and was approved by the Medical Ethics Committee of Medisch Spectrum Twente, Enschede, the Netherlands. All participants gave written informed consent before participating in the study.

2.2 APPARATUS
Multi-joint coordination was investigated using a custom made device (Motekforce Link, Culemborg, The Netherlands), the Double Inverted Pendulum Perturbator (DIPP) (Engelhart et al., 2015). The DIPP consists of two manipulators and applies forces using push/pull rods at hip and shoulder level (Figure 1). Both manipulators were adjustable in height to align the rods to the participant’s hip and shoulder level. The manipulators were force controlled such that force disturbances and force fields
could be simultaneously applied. Force disturbances are pushes and pulls on the human body, required for identification of the neuromuscular controller. Force fields were applied at hip level to evoke adaptation of multi-joint coordination. Participants experienced the force fields as if they were attached to a spring with varying stiffness, pushing or pulling them back to an equilibrium position. During the experiments, participants wore a safety harness to prevent falling. The harness did not constrain movements or provide support in any way.

2.3 DISTURBANCE SIGNALS
The force disturbances were unpredictable and continuous multisine signals in a range from 0.05 – 5 Hz (Figure 2). The shoulder and pelvis disturbances were zippered multisine (Pintelon and Schoukens, 2012), such that each disturbance contained nine excited frequencies at an interleaved frequency grid. Both multisine signals had a period of 20 seconds, and were repeated nine times over a time course of three minutes. Disturbances had a peak-to-peak amplitude of 80 N.

2.4 PROCEDURES
Participants were attached to the DIPP and stood with eyes open, without shoes, and their arms folded across their chest. Participants were instructed to maintain a normal upright stance position. Force disturbances at the hip and shoulder were applied simultaneously. External force fields were applied only at the level of the hip. The (spring) stiffness of the force field was set such that it compensated partially for the gravitational stiffness around the participant’s ankle joint, and was normalized to $mgl$ (in which $m$ is the mass of the participant, $g$ the gravitational acceleration (9.81 m/s$^2$) and $l$ the height of the CoM (Center of Mass), estimated by $0.575*\text{height of the participant}$).

Within the five experimental trials, a fixed stiffness level was set, expressed in a percentage of full compensation. One baseline trial was recorded without force field (0%). Three force fields were stabilizing the human body (20%, 50%, and 80%), one force field destabilized the human body (-20%). All trials were randomized, and participants were allowed to rest between trials according to their needs. Prior to the experiments, participants were allowed to get familiarized with the disturbances according to their individual needs. In case of the five force fields trials, participants were allowed to
familiarize approximately 10 seconds before recording the data. This familiarization reduced any transient effects in the responses.

2.5 RECORDINGS

Body kinematics were measured using two draw-wire potentiometers (Celesto SP2-25, Celesto, Chatsworth, CA, United States), which were attached to the end of the push/pull rods and measured the displacement of the participant’s upper and lower body segments. A dual-force plate (AMTI; Watertown, USA), measured the ground reaction forces and torques in six degrees of freedom under each foot. All signals were recorded with a sample frequency of 1000 Hz and processed in Matlab (The MathWorks, Natick, MA, United States).

2.6 DATA ANALYSIS

To study the age-related changes in adaptation of multi-joint coordination, different data-analysis methods were used. In various studies (Horak and Nashner, 1986; Creath et al., 2005), multi-joint postural control is described by the ankle and hip strategy, which are defined as movements around the ankle and hip joints. Therefore, sway and torque responses of young and elderly participants were compared, expressed by the Root Mean Square (RMS). The RMS gives an indication of the effective amplitude, instead of a maximum, yielding a representative value for the response amplitudes. Secondly, covariance descriptors (Kuo et al., 1998) resulted from a kinematic analysis, which included postural coordination between the ankle and hip joints. To include multi-segmental influences in postural control (Boonstra et al., 2013), system identification techniques were used, which resulted in Frequency Response Functions (FRFs). The FRFs describe the relationship between the torques and the angles as a response to the disturbances only, i.e. inter-segmental coordination of the neuromuscular controller is identified. Finally, to find physiologically relevant parameters that describe the underlying mechanisms of multi-joint coordination, a model was fit to the FRF. This resulted in an estimate of time delays and parameters describing intrinsic and reflexive feedback properties.
**Preprocessing**

All signals were filtered with a phase preserving fourth-order Butterworth filter, with a cutoff frequency of 10 Hz. The ankle and hip angle were calculated with goniometric rules from the recorded upper and lower body displacements and the participant’s dimensions. The ankle angle was described as the angle of the lower body with respect to the foot (and thus the horizontal, as the support surface was fixed). The hip angle was described as the angle of the upper body with respect to the longitudinal axis of the upper leg. From the force plate data, the total ankle and hip torques were calculated with inverse dynamics (Winter, 1990; Schwab, 1998). Data were segmented in 20 s periods of the disturbance signals, yielding nine data blocks per trial. From each data block, the mean and trends were removed. The responses were averaged over the nine data blocks for each condition and participant and used for further analysis.

**Root Mean Square and Covariance Descriptor**

A description of the response amplitude of the ankle and hip angle and the ankle and hip torque was given by the RMS.

To represent the upper and lower body movement and their coupling, the two-by-two covariance matrix of the ankle and hip angle was used (Kuo et al., 1998).

\[
Q = \text{cov}(\theta) = \frac{1}{n - 1} \sum_{i=1}^{n} (\theta_i - \bar{\theta})(\theta_i - \bar{\theta})^T
\]

In which \(\theta_i\) is the \(i^{th}\) time sample of the vector containing the ankle and hip angles. The diagonal terms of the covariance matrix represent the variances of the ankle and hip angle. The off-diagonal terms represent the inter-segmental coupling between the ankle and hip.

The covariance matrix may be described by an ellipse (Kuo et al., 1998; Speers et al., 2002). The extent of the ellipse along the horizontal and vertical axes is proportional to the RMS motion of the ankle and hip, respectively. The eigenvalues of the covariance matrix, \(\lambda_1\) and \(\lambda_2\), describe the squared lengths of the major and minor ellipse axes, respectively. The long axis of the ellipse (\(\lambda_1\))
represents the amount of hip strategy utilized by the participant, while the minor axis of the ellipse $(\lambda_2)$ is an approximate indicator of the amount of CoM movement. The eigenvector with the largest eigenvalue was used to calculate the orientation angle ($\alpha$) of the ellipse. The angle quantifies the direction of the relationship between the ankle and hip angles.

**FREQUENCY RESPONSE FUNCTIONS**

The dynamical properties of the neuromuscular controller were quantified in the frequency domain by FRFs. The FRFs consist of two parts: a magnitude and a phase, describing the relationship between disturbances and the responses in terms of magnitude and time, respectively. Using closed-loop system identification techniques (van der Kooij et al., 2005; Engelhart et al., 2015), the FRFs of the neuromuscular controller ($H_c$) are computed from the experimental data, according to:

$$H_c = -S_{dT}(S_{d\theta})^{-1}$$

In which $S_{dT}$ and $S_{d\theta}$ are the cross-spectral density matrices between the external disturbances ($d$) to the corrective ankle and hip torques ($T$) and the ankle and hip angles ($\theta$), respectively, resulting in a two-by-two FRF matrix ($H_c$). Inverting the $S_{d\theta}$ matrix and multiplying it with the $S_{dT}$ matrix requires that the matrix components are known at all excited frequencies, which is untrue for the zippered multisine, as both disturbance signals contain different frequencies. Therefore, the complex numbers of the cross-spectral densities were interpolated in terms of magnitude and phase, to obtain all matrix components for the full range of excited frequencies in the zippered multisine. The FRFs were only evaluated at the frequencies where the disturbance signals contained power. The FRFs were normalized for the gravitational stiffness $(mgl$: $m$: total body mass, $l$: CoM height and $g$: gravitational acceleration) to compensate for differences in the participants’ mass and pendulum height, which influences the FRFs.

$H_c$ consists of two direct terms, covering the FRFs from ankle angle to ankle torque ($H_{c\theta ank} T_{ank}$) and from hip angle to hip torque ($H_{c\theta hip} T_{hip}$). These direct terms quantify the ankle and hip contributions to balance control. Furthermore, there are two indirect terms, which cover the FRFs
from ankle angle to hip torque \((H_{c,\theta_{ank}2T_{hip}})\) and from hip angle to ankle torque \((H_{c,\theta_{hip}2T_{ank}})\) and reflect the intersegmental coupling (Boonstra et al., 2013).

**MODEL DESCRIPTION**

A FRF describes the behavior of the system, but it does not reveal which physiological mechanisms are underlying the system. To relate the changes in behavior to the changes in the underlying physiology, a model of the neuromuscular controller was fitted to the non-normalized FRFs (Figure 3).

The neuromuscular controller stabilizes the human body by generation of joint torques. The corrective joint torques around the ankles and hips result from intrinsic feedback together with delayed neural feedback. Each system in the neuromuscular controller was described by a mathematical formula (i.e. transfer function) with parameters describing the physiology. This resulted in a model for the four terms of the neuromuscular controller:

\[
H_{c,\theta_{ank}2T_{ank}} = H_{p}^{ank} + H_{r}^{ank2T_{ank}}H_{TD}^{ank}
\]

\[
H_{c,\theta_{hip}2T_{ank}} = H_{r}^{hip2T_{ank}}H_{TD}^{ankhip}
\]

\[
H_{c,\theta_{ank}2T_{hip}} = H_{r}^{ank2T_{hip}}H_{TD}^{ankhip}
\]

\[
H_{c,\theta_{hip}2T_{hip}} = H_{p}^{hip} + H_{r}^{hip2T_{hip}}H_{TD}^{hip}
\]

The transfer function of intrinsic feedback \((H_{p})\) describes the muscle and tendon dynamics together with the soft tissue properties. Intrinsic feedback is described for each joint by only stiffness \((H_{p} = K_{p})\). Intrinsic dynamics only exists directly at joint level, i.e. the direct terms of the neuromuscular controller, with different values for the ankle and hips \((H_{p}^{ank}, H_{p}^{hip})\).

In the transfer function of the lumped time delay \((H_{TD} = e^{-\tau_d s})\), \(\tau_d\) represents the sum of neural conduction time (transport delay), an electromechanical delay (to activate the muscles), and the
processing time of sensory information. A separate lumped delay was introduced for each path length the sensory information travels, e.g. the direct terms of the neuromuscular controller had separate delays \( H_{TD}^{ank}, H_{TD}^{hip} \) and the indirect terms of the FRF had equal delays \( H_{TD}^{ank,hip} \).

The transfer function of reflexive feedback \( H_r \) was represented by a matrix with stiffness and damping terms, relating the joint torques to the ankle and hip angles and angular velocities. This resulted in four transfer functions.

\[
H_r^{ank2T_{ank}} = K_{ank2T_{ank}} + D_{ank2T_{ank}} S
\]

\[
H_r^{hip2T_{ank}} = K_{hip2T_{ank}} + D_{hip2T_{ank}} S
\]

\[
H_r^{ank2T_{hip}} = K_{ank2T_{hip}} + D_{ank2T_{hip}} S
\]

\[
H_r^{hip2T_{hip}} = K_{hip2T_{hip}} + D_{hip2T_{hip}} S
\]

The model had a total of 13 physiologically interpretable parameters. Table 2 shows the model parameters and how they are estimated, as is outlined next.

**MODEL FITTING**

The transfer functions from the model were fitted to the experimental non-normalized FRFs using a nonlinear least-squares optimization algorithm. The algorithm searches for a parameter set that minimizes the objective function \( J_{i,f_k} \).

\[
J_{i,f_k} = \left( \sum_{i=1}^{4} \sum_{k=1}^{18} J_{i,f_k} \right)
\]

\[
SSE = \left( \sum_{i=1}^{4} \sum_{k=1}^{18} J_{i,f_k} \right)
\]

The logarithmic difference between the FRF \( H_c \) based on the calculated parameter vector \( p \) and the estimated FRF \( \tilde{H}_c \) obtained from experimental data was summed over the frequencies \( f_k \),
The objective function is chosen such that there is more emphasis on the low frequencies \((\frac{1}{1+f_k})\), where stiffness typically manifests. In addition, a relative error is calculated for all frequencies on a logarithmic scale.

The optimization algorithm was run 20 times with random initial conditions to assure a global minimum was found. The best parameter set was obtained from the fit with the lowest SSE value. In addition, the goodness of fit (GOF) describes how well the data compare to the estimation with the parameter set and is expressed as:

\[
\text{GOF}_i = 1 - \frac{\sum |h_{ci}(f_k,p) - \tilde{h}_{ci}(f_k)|^2}{\sum |h_{ci}(f_k)|^2} \times 100\%
\]

The models were fitted for each participant on all five experimental trials simultaneously. To limit the number of parameter combinations in the fit, lumped time delays and intrinsic stiffness were constant over trials in which force field levels were varied. The lumped time delays of the indirect terms were calculated as the averages of the direct terms. Reflexive feedback was variable over trials as it was assumed that subjects alter their control action when they are exposed to a force field.

After the model fitting procedures, the passive stiffness and the reflexive stiffness and damping values were normalized for the gravitational stiffness of each participant, to correct for the differences in mass and height between participants.

**Statistical Analysis**

The characteristics of the participants were represented by mean and standard deviation in case of a Gaussian distribution. Else, median, and inter quartile range or number and percentage were presented. To test significant differences between groups, an independent two-sided t-test was performed. In case of the non-normal distributed values, a Mann-Whitney U test was performed.

To test significant differences in RMS and covariance descriptors between age and force field levels, linear mixed models were used. Age and force field level were fixed effects. To account for the
repeated measurements, participant intercept was included as a random effect. In addition, interaction effects between age and force field levels were studied.

For statistical analysis of the FRFs, the magnitude of each FRF was logarithmically transformed to make the data normally distributed. Subsequently the magnitudes were averaged within three frequency bands (<1 Hz, 1-2.5 Hz and 2.6-5 Hz). The lowest frequencies generally describe the stiffness properties of the system, whereas the magnitude at middle and high frequencies is shaped by damping and mass (inertia), respectively. Linear mixed models were used to test significant differences in FRFs between age and force field levels for each frequency band. Age group, force field level, and frequency band were fixed effects and participant intercept was a random effect. In addition, interaction effects between age group and force field level were studied.

To test significant differences of the estimated model parameters that were constant over trials, an independent two-sided t-test was performed, as there can only be an effect of age and not of force field level. To test significant differences between age and different force field levels in the estimated model parameters that were allowed to vary over trials, linear mixed models were used. Age group and force field level were fixed effects, and participant intercept was a random effect. In addition, interaction effects between age and force field level were studied.

For all tests, the significance level ($\alpha$) was set at 0.05. All analyses were performed with SPSS version 22.0 (SPSS, Chicago, IL).

### 3. RESULTS

Characteristics of the group of healthy young and healthy old participants are presented in Table 1 to illustrate the health of the study population. The elderly used more medication ($p=0.001$), had lower MMSE score ($p=0.023$), and lower handgrip strength ($p=0.014$) compared with the young participants (Table 1). All subjects were able to maintain balance during the disturbances and the force field trials.
3.1 Root Mean Square

Figure 4 shows the joint angles and torques as a response to the force disturbances for a representative young and elderly participant (without application of a force field). Figure 5 shows the RMS of the ankle and hip angle and the ankle and hip torque. The elderly had smaller ankle (p=0.001) and hip (p=0.003) angles than the young participants. The exerted ankle and hip torques were not found to be significantly different between the two groups.

For every 10% increase in external force field, the averaged postural responses decreased, namely the ankle angle with -0.02 deg (p<0.001), the hip angle with -0.03 deg (p<0.001), the ankle torque with -0.7 Nm (p<0.001) and the hip torque with -0.07 Nm (p<0.001).

An interaction effect was found between age and force field level; with increasing force field level, the decrease in ankle (p=0.007) and hip (p=0.004) angle was less in the elderly than the observed decrease in young participants. No interaction effect was found between age and force field level for the ankle and hip torque. The elderly and the young participants showed comparable adaptation of their joint torques to different force field levels.

3.2 Covariance Descriptor

Figure 6 shows the covariance descriptor for the young and elderly participants for the different force field levels. The length of the ellipse major axis was lower in the elderly compared to the young participants (p=0.013). No differences were found for the minor axis or the orientation of the ellipse between the young and elderly.

When applying a stiffer force field, the orientation angle of the ellipse increased and the covariance descriptors decreased. For every 10% increase in force field, the values (averaged over the repetitions of the disturbance signals) altered; \( \lambda_1 \) (-0.013, p<0.001), \( \lambda_2 \) (-0.001, p<0.001) and \( \alpha \) (0.002 rad, p<0.001).
An interaction effect between age and force field level was found; the decrease of $\lambda_1$ was less in the elderly compared to the young ($p=0.01$). No interaction effect was found between age and force field level for the $\lambda_2$ and $\alpha$; the elderly adjusted these values compared to the young participants.

### 3.3 Frequency Response Functions

Figure 7 shows the neuromuscular controller FRF ($\tilde{H}_c$) of the young and elderly in the baseline trial, when only disturbances were applied without external stiffness (0%).

There was a main effect of age, as the magnitude of $H_{c,\theta_{ank2T_{ank}}}$ was higher in the elderly for the lowest ($p=0.002$) and midrange ($p<0.001$) frequencies. For the other terms, the magnitude was higher in the elderly in the midrange frequencies of $H_{c,\theta_{hip2T_{ank}}}$ ($p=0.001$) and $H_{c,\theta_{ank2T_{hip}}}$ ($p=0.006$). No significant differences with age were found in any of the frequency bands for $H_{c,\theta_{hip2T_{hip}}}$ and no differences were found for the high-frequency range of the neuromuscular controller.

When applying a force field, the FRF magnitude changed (Figure 8). With increasing force field, the FRF magnitudes became lower for the entire frequency range in $H_{c,\theta_{ank2T_{ank}}}$ ($p<0.001$, for all frequency bands) and $H_{c,\theta_{hip2T_{ank}}}$ ($p<0.001$, for all frequency bands). A significant decrease of FRF magnitude was also found for the lowest and midrange frequencies in $H_{c,\theta_{ank2T_{hip}}}$ ($p<0.005$, for both frequency bands) and the midrange frequencies of $H_{c,\theta_{hip2T_{hip}}}$ ($p<0.016$).

An interaction effect between age and force field level was found for the low and midrange frequencies in $H_{c,\theta_{ank2T_{ank}}}$ ($p=0.027$ and $p=0.008$) and $H_{c,\theta_{hip2T_{ank}}}$ ($p=0.026$ and $p=0.002$). Elderly participants reduced the FRF magnitude around the ankle less compared to young participants. The FRF magnitude adjustment around the hips was comparable in both groups. For example, the magnitude of $H_{c,\theta_{ank2T_{ank}}}$ for the lowest frequencies in young participants was decreased by 21% for the stiffest force field (80%), whereas for the elderly it was decreased by 14%. For the destabilizing
force field (-20%), postural responses increased 28% in young, whereas for the elderly they were only increased by 8%.

3.4 MODEL PARAMETERS

The GOF values were averaged (± standard deviation) over force field levels and participants, resulting in a GOF value for each FRF term ($H_{c,\theta_{ank}^2T_{ank}}$, $H_{c,\theta_{hip}^2T_{ank}}$, $H_{c,\theta_{ank}^2T_{hip}}$ and $H_{c,\theta_{hip}^2T_{hip}}$).

The average GOF for the young participants were 81 ± 14%, 70 ± 18%, 50 ± 23% and 75 ± 17% and for the elderly were 75 ± 13%, 70 ± 13%, 63 ± 20% and 82 ± 14%. Similar GOF values were obtained for the young and elderly. Different GOF values were found between conditions, e.g. some of the conditions had higher GOF values, and thereby were fit better than others. Furthermore, the GOF values of the direct FRF terms were higher than those of the indirect terms and the GOF values of $H_{c,\theta_{ank}^2T_{hip}}$ was lowest. The goodness of fit is also displayed in Figure 9, which shows the estimated FRFs together with the model fit for a representative participant.

Figure 10 shows the estimated parameters for the young and elderly participants. The elderly had larger lumped time delays for the hips (p<0.001) compared to the young participants. The lumped time delays of the ankle were not found to be significantly different between age groups. Furthermore, the intrinsic ankle stiffness (p=0.007) was higher in elderly, but no significant differences were found for the intrinsic hip stiffness. The reflexive stiffness $K_{ank}^2T_{ank}$ was significantly higher in elderly than in young participants (p=0.036). None of the other reflexive stiffness and damping terms were found significantly different between young and elderly.

When increasing the force field level in both groups, all reflex stiffness and damping values decreased (p<0.04), except for the reflexive stiffness around the hip; $K_{ank}^2T_{hip}$ and $K_{hip}^2T_{hip}$. An interaction effect was found between age and force field level only for $K_{ank}^2T_{ank}$ (p=0.04). The elderly reduced their reflexive stiffness around the ankle joint less for increasing force fields compared to the young.
4. DISCUSSION

The results of this study show age-related differences in multi-joint coordination. The elderly swayed less than the young participants, and the elderly showed a reduced hip strategy. As the corrective joint torques were not significantly different between age groups, the FRF magnitude was higher in the elderly. The ratio of the corrective joint torques and the body sway was increased, i.e. the elderly exhibit a higher stiffness. Parameter estimation showed that the elderly have higher intrinsic stiffness and reflexive stiffness around the ankle joint. When an external force field was applied, both age groups lowered their postural responses, expressed as lower FRF magnitude around the ankle. However, the elderly adapted their postural responses less compared to the young participants.

4.1 PARTICIPANT CHARACTERISTICS

Based on the inclusion criteria, all young and elderly participants were characterized as healthy. None of the young participants were taking medication. The elderly used medication, but none of the medication had a known influence on balance control. Comparison between young and elderly participants showed a significantly lower MMSE score in the elderly. It is known that cognitive processing has an influence on the control of balance (Teasdale and Simoneau, 2001; Doumas et al., 2009; Ambrose et al., 2013; Stijntjes et al., 2015), e.g. low cognitive function increases the risk of imbalance. Although all of the elderly were characterized as healthy with normal cognitive function, the lower MMSE score might have influenced the results. None of the participants reported fear of falling or fatigue during the experiment.

4.2 ROOT MEAN SQUARE AND COVARIANCE DESCRIPTOR

Results show that the elderly swayed less when being disturbed by forces at the hip and shoulder, compared to the young participants. These results were in contrast with multiple studies of quiet stance, showing that body sway increased with age (Abrahamová and Hlavačka, 2008; Demura et al., 2008; Pasma et al., 2014a). In our study, the balance control system was externally disturbed. Humans altered their feedback gains to correct for these disturbances, i.e. the amount of joint
torque relative to the amount of joint motion was adjusted. To be more specific, with increasing stabilizing force field level, the feedback gains were reduced. In quiet stance, the balance control system is mainly influenced by internal disturbances, such as sensory and motor noise, which cannot be corrected for. Altering feedback gains during quiet stance would amplify the internal disturbances, which increases body sway (Speers et al., 2002). This could explain why in quiet stance, the elderly sway more and that it is possible that in perturbed stance the elderly sway less.

Covariance descriptors were used to describe not only whether the application of force fields resulted in changes of sway, but also in postural coordination. Eigenvectors and eigenvalues were used to describe independent combinations of joint movements, defined by principle component analysis (Kuo et al., 1998; Alexandrov et al., 2001; Hsu et al., 2007). From the covariance matrix of the ankle and hip angle, the first component $\lambda_1$ may be interpreted as double-inverted pendulum behaviour and quantifies the amount of hip strategy. Our results show that $\lambda_1$ is smaller in the elderly, indicating they have less hip strategy and behave more like a single inverted pendulum. The second component $\lambda_2$ was not found to be significantly different between age groups, indicating the control of the CoM was similar. In a previous study (Hsu et al., 2013) where external disturbances (platform translations) were applied, the elderly exhibited a less flexible stance and a decreased covariance between the joints, while the CoM excursion was not significantly different between the groups. This is in accordance with our findings.

When a stabilizing force field was applied, a decrease of both $\lambda_1$ and $\lambda_2$ was found as maintaining standing balance became easier. A decrease in $\lambda_1$ indicates that the combination of opposing ankle and hip motion was reduced; i.e. hip strategy became less with increasing force field. A decreased $\lambda_2$ value, indicates that a reduction was found in the combination of movements, dominated more by the ankle angle. The negative relation remained; a positive ankle angle was accompanied with negative hip angle. There was an interaction effect between age and force field level. The elderly
adapted their hip strategy less than young participants. This might be explained by the fact that the
\( \lambda_1 \) was lower in elderly at baseline. The elderly adapted their control of CoM similar to the young.

4.3 FREQUENCY RESPONSE FUNCTIONS

Measured joint angles contain not only a sway response due to the disturbances, but also contain the
subject's own spontaneous body sway (remnant sway due to e.g. motor and sensory noise (van der
Kooij and Peterka, 2011)). As both RMS measures and covariance descriptors are based directly on
these measured joint angles, they do not distinguish between the responses due to the disturbances
and spontaneous sway. Therefore, in this study we used system identification techniques to solely
identify the neuromuscular control mechanisms from the closed-loop feedback system, i.e. the
corrective joint torques as a response to body movement around the joints. The neuromuscular
controller dynamics and the coordination between joints has frequency specific effects, which were
shown in FRFs. Compared to RMS values and covariance descriptors, the FRF is potentially a more
informative measure. The FRF describes only the part of the angles and corresponding corrective
joint torques as a response to the disturbances, i.e. changes in the neuromuscular controller only.

In the elderly, the RMS values of the joint angles were lower and the joint torques were comparable
to young adults. These results were also seen in higher FRF magnitude of elderly, at the lowest
frequencies of the direct ankle term. Stiffness is assumed to dominate the magnitude of the FRFs in
the low-frequency range, indicating that the elderly have higher ankle joint stiffness. When the pull
of gravity is compensated for by an external force field, the elderly adapted their FRF magnitudes
around the ankle less compared to young. These age-related differences in adaptation to force field
levels were also found in the RMS outcomes and the covariance descriptors.

In addition to the adaptation at the lowest frequency range, also in the midrange frequencies
significant differences were found between force field conditions. The midrange frequencies are
mostly affected by damping properties. Increasing damping reduces oscillations in the response to
external disturbances (Cenciarini, 2010). No significant differences were found between force fields
conditions at high frequencies, because the FRF magnitude at high frequencies is generally shaped by
the mass properties of the participants, which were not significantly different between young and elderly (Table 1).

4.4 MODEL PARAMETERS

Estimating model parameters on the FRF reveals the underlying factors of the neuromuscular controller. The elderly exhibit a larger intrinsic stiffness and reflexive stiffness of the ankle compared to the young, which is in concordance with our expectations following the results of the FRFs. When the force field level was increased, both the young and elderly reduced their reflexive stiffness and damping. However, the reflexive feedback gains around the ankle joint were reduced less in elderly compared to young participants.

The body is mostly represented as an inverted pendulum, based on body rotation around the ankle joint. Our study includes a hip joint, which might result in different values of the estimated parameters compared to other studies. Kiemel et al. (Kiemel et al., 2008) estimated intrinsic joint parameters of both the ankles and hips using system identification techniques based on electromyography (EMG) signals and joint angles, when healthy young subjects were faced with visual scene disturbances. The intrinsic stiffness of the ankle and hips were found to be 293 Nm/rad and 95 Nm/rad, respectively. Cenciarini et al. (Cenciarini, 2010) found intrinsic ankle stiffness of 157 and 99 Nm/rad for the young and elderly, respectively, when exposed to support surface tilts. The estimated stiffness values (i.e. the non-normalized values) in this study are in the same range as the other two studies; however in our study the elderly exhibit larger intrinsic ankle stiffness compared to the young.

Estimated reflexive stiffness and damping of the ankle in the current study are within the ranges earlier found in the literature varying from 898 - 1500 Nm/rad and from 288 - 480 Nms/rad (Peterka, 2002; Mahboobin et al., 2005; Cenciarini, 2010; Davidson et al., 2011). Also others found a comparable reflexive stiffness between young and elderly, which is in accordance with our results (Ho and Bendrups, 2002). Upper body stiffness and damping between 100-300 Nm/rad and 20-60
Nms/rad was found in healthy young subjects (Goodworth and Peterka, 2012), being similar to our results.

The time delays as estimated in this study consisted of processing time, electromechanical delay, and neural conduction time. Previous studies, in which the human body was represented as an inverted pendulum, found a time delay of approximately 172 ms, which was not significantly different between the young and elderly (Cenciarini, 2010). Other studies found time delays in the range of 100-200 ms (Peterka, 2002; Mahboobin, 2007; Davidson et al., 2011), and the elderly exhibited significantly higher delays, compared to the young (Davidson et al., 2011). This compares to our results.

4.5 UNDERLYING MECHANISMS

In this study we used system identification techniques to quantify age-related differences in adaptation of multi-joint coordination. With current clinical balance tests, the subsystem that is responsible for the observed behavior, remains largely unknown (Visser et al., 2008; Pasma et al., 2014b). With system identification techniques and parameter estimation, the contribution of the underlying subsystems can be unraveled, as the balance control mechanism is expressed in physiologically relevant parameters (Engelhart et al., 2014). Identification of the factors that contribute to altered postural responses in the elderly may provide insight into the mechanisms resulting in impaired balance and finally falling. More insight in these factors and mechanisms, might help to develop and test targeted interventions to reduce the risk of falling. For example, a higher joint stiffness is found, which can increase the risk of falling in elderly (Ishida et al., 2008). Although it is currently unclear what causes the increased stiffness, it could result from e.g. altered muscle properties or could indicate compensatory co-contraction for increased noise.

This study showed a higher intrinsic stiffness of the ankle in elderly compared with young, which could be explained by age-related changes in fiber elasticity in type I and IIa muscle fibers. This change in elasticity will result in a higher stiffness of the muscle (Ochala et al., 2007). On the other
hand, a higher stiffness could be a way to compensate for the deterioration of sensory systems. Previous studies showed that the proprioceptive information from the distal joints (i.e. the ankle) was insufficient to maintain standing balance due to peripheral neuropathy (Horak et al., 1990). A higher intrinsic stiffness, for example generated by co-contraction (Benjuya et al., 2004), would increase the response of the proprioception and therefore will enhance the proprioceptive input from the ankle (Accornero et al., 1997).

Furthermore, we found significant differences in reflexive stiffness around the ankle joint between young and elderly. Besides, elderly adapt their reflexive stiffness around the ankle less compared with young, when the force field at hip level was increased resulting in a more stable ankle joint. This difference might be due to the loss of flexibility to control multiple joints (Ooteghem, 2009; Hsu, 2013). It could be that age-related changes both in muscles (e.g. intrinsic stiffness and muscle strength) and sensory systems (e.g. peripheral sensory loss and afferent and efferent conduction impairment) results in less options to control balance.

Last, the time delays of the hip were higher in elderly compared with young. The time delays estimated in this study comprised of processing time, electromechanical delay, and neural conduction time. Therefore, a higher time delay might be due to slowing down of the nerve conduction speed in afferent and efferent pathways with age due to a decrease in the number of neurons, loss of myelination and other neural changes (Sturnieks et al., 2008; Barin and Dodson, 2011). On the other hand, a higher time delay with age, can be due to deficits in stimulus encoding, central processing and response initiation (Horak et al., 1989).

### 4.6 STRENGTHS AND LIMITATIONS

In contrast with current RMS values and covariance descriptors, system identification techniques make it possible to derive parameters of the neuromuscular controller in multi-segmental balance control. Nevertheless, the results might be influenced by various factors. First, the control scheme of the neuromuscular controller may not be an adequate representation of actual postural control, and
it may not describe all age-related changes in standing balance. We assumed that the states of the human body (joint angles and angular velocities) were fully known and the sensory information was “perfect”. However, with age, the sensory systems might become impaired (Sturnieks et al., 2008). Furthermore, we did not model the integration of sensory information, i.e. the process of sensory reweighting (Peterka and Loughlin, 2004; Mahboobin et al., 2009). Therefore, age-related changes in the quality of sensory information and sensory reweighting capacities were not studied.

Second, estimation of model parameters based on experimental data is a constant trade-off between a good model fit and the least amount of parameters. Increasing the number of parameters improves the fit; however the physiological interpretation becomes difficult. Two aspects are important when interpreting a parameter set obtained from a model fit on experimental data. The first is independency, i.e. when two parameters give similar contributions to the FRF, redundancy exists, which hampers the physiological interpretation. Secondly, identifiability, meaning that a parameter has to contribute to the FRF within the excited frequency band in order to assure that the influence of that parameter can be detected. The goal of model fitting is to find the most compact model with parameters that are both independent and identifiable.

During the process of model fitting, various combinations of parameter sets were estimated and validated, based on previously used models in literature (Kiemel et al., 2008; Goodworth and Peterka, 2012). In our fitting procedure, the intrinsic damping properties and the activation dynamics (mapping of EMG signals to joint torques) were found unidentifiable. Muscle-activation dynamics did not influence the magnitude of the FRF in the frequency range the data was obtained (0.05-5Hz). However, the FRF phase was affected. Interaction existed with the lumped delays, which also shaped the FRF phase, which resulted in poorly interpretable values. We therefore decided not to include the muscle-activation dynamics in the model, and we attributed all phase changes to the lumped delays (which represents the sum of transport delay, the processing time of sensory information and an electromechanical delay). Interaction also existed in the direct terms of the FRF \( H_c \theta_m k 2T_m k \) and
between the intrinsic stiffness, the reflexive stiffness, and the lumped delay. The FRF magnitude displayed the sum of the stiffness values. The identified FRF phase was somewhere between the phase of 0 (due to the intrinsic stiffness) and the phase due to the lumped delay. The result is that only the product of reflexive stiffness and time delay was identifiable. To solve the redundancy among the three parameters, we assumed two of the three interacting parameters to remain constant over the five experimental conditions. The changes in intrinsic properties and lumped delays were expected to be much smaller than the changes in reflexive properties (Peterka, 2002; Cenciarini, 2010). Therefore, the intrinsic properties and lumped delays were averaged over conditions, and the reflexive stiffness was left to vary over conditions.

However, the assumption of average intrinsic stiffness over the experimental conditions might have influenced our results. Intrinsic stiffness scales with the contraction level of the muscles, which is influenced by co-contraction or other external factors (such as a force field) (Ludvig et al., 2011). With increasing force field levels, the pull of gravity on the participants’ CoM was reduced, and this possibly also reduced the activation level of the muscles as control of balance became easier. With decreased muscle activation, the intrinsic stiffness will also decrease. Due to our assumption, variations in intrinsic stiffness were now captured in the reflexive stiffness values.

As mentioned before, the lumped delay in the model was estimated from the FRF phase, which resulted from the contribution of intrinsic stiffness (which acts without a delay and has a phase shift of zero) and reflexive stiffness (which acts with a delay and induced a negative phase shift). If there was an increased contribution of corrective torque due to intrinsic properties compared with reflexive feedback, the estimated lumped delay might appear to decrease (Peterka, 2002). Therefore the lumped delay parameter in the model can be better thought of as an “effective delay” rather than as a parameter representing actual delays in neural processing, transmission, and muscle activation. In case one is interested in the separate contribution of the transport delay, the
electromechanical delay and the processing time in the lumped delay, electromyography measures can be of additional value.

5. **CONCLUSION**

In this study we used novel system identification techniques to derive a description of the neuromuscular control mechanisms in multi-joint balance control, by applying force disturbances at the level of the hip and between the shoulder blades. Adaptation of multi-joint coordination was induced by external force fields, represented as virtual springs at the pelvis with various stiffness levels. Our results demonstrate that humans adapt to force fields by altering their postural responses, i.e. reflexive stiffness and damping. However, elderly adapted their reflexive stiffness around the ankle less compared to young participants. In addition to higher reflexive ankle stiffness, elderly had higher intrinsic ankle stiffness and larger lumped time delays of the hip. Insight in the factors that alter multi-joint coordination with age, could provide insight into the mechanisms that influence fall risk. Adaptability of multi-joint coordination is an essential feature of standing balance control, in order to adjust to various conditions. Reduced ability to adapt may underlie increased risk of falls with ageing.
6. ACKNOWLEDGEMENTS

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Table 1: Participant characteristics. All parameters are presented as mean with standard deviation unless indicated otherwise. Abbreviations: IQR, interquartile range; MMSE, Mini-Mental State Examination; SPPB, Short Physical Performance Battery.

Table 2: Overview of the parameters as used in the model fits. The parameters were fit to the five force field conditions simultaneously and for each subject individually. Time delay and intrinsic feedback parameters were kept constant over the five conditions, while the values for the reflexive feedback parameters varied over the conditions. The time delay for ankle-hip and vice versa, were calculated as the average of the time delay ankle and time delay hip.

Figure 1: Double Inverted Pendulum Perturbator (DIPP) with a subject standing on a force plate (A) and attached to two manipulators at shoulder and hip level (B). Both manipulators are adjustable in height and driven by an electromotor, pushing and pulling the subject. Body kinematics are measured by potentiometers attached to the rod (C). A safety harness is attached to the pyramidal construction (D) and an emergency button is mounted on the frame (E).

Figure 2: Time series of the applied force disturbances at the pelvis and between the shoulder blades, with the corresponding Power Spectral Density (PSD). The force disturbances are zippered multisine, as the excited frequencies (indicated with circles) are independent.

Figure 3: Schematics of the human balance control system. The human body was represented as a double inverted pendulum, with disturbances acting at the hip and shoulder level. The model of the neuromuscular controller was used for parameter estimation. The inputs were the ankle and hip angle ($\theta_{\text{ank}}, \theta_{\text{hip}}$) and the outputs the corrective joint torques ($T_{\text{ank}}, T_{\text{hip}}$). Intrinsic dynamics were modelled as a spring and were different for the ankle and hip joints. Reflexive control and time delay dynamics were MIMO transfer functions (shown as dotted boxes), in which interaction existed between the ankle and hip joint signals.
Figure 4: Joint angles and joint torques in response to the applied disturbances of a representative young (left) and elderly (right) participant in the baseline trial without force field (0%). The average over the nine disturbance cycles is indicated with the black line; the grey area represents the standard deviation.

Figure 5: Root Mean Square (RMS) values of the joint angles and torques for young and elderly participants per force field level represented by mean and standard deviation. The asterisks represent significant differences with age (*), force field (**) or the interaction between age and force field (***)

Figure 6: Covariance descriptor of the young and elderly participants per force field level. $\lambda_1$ and $\lambda_2$, describe the squared lengths of the major and minor ellipse axes, and the orientation of the ellipse is given by $\alpha$.

Figure 7: Baseline differences in normalized Frequency Response Function (FRF) between young (black) and elderly (grey) participants, where only the two disturbances were applied, without additional force fields (0%), represented by mean and standard deviation. The FRF consists of four terms, two direct terms from ankle angle to ankle torque ($\theta_{\text{ank}}^\times T_{\text{ank}}$) and from hip angle to hip torque ($\theta_{\text{hip}}^\times T_{\text{hip}}$), and two indirect terms from hip angle to ankle torque ($\theta_{\text{hip}}^\times T_{\text{ank}}$), and from ankle angle to hip torque ($\theta_{\text{ank}}^\times T_{\text{hip}}$). The asterisks represent the frequency bins in which there is a significant difference with age (*).

Figure 8: Adaptation of normalized Frequency Response Function (FRF) magnitude in the young (upper part) and elderly (lower part) participants. For each force field level the mean is shown. For displaying reasons only the standard deviation was shown for the baseline trial. Standard deviations of the other force field levels were comparable to the baseline trials. The asterisks represent the frequency bins in which there are significant differences with force field (**) or the interaction between age and force field (***)

Figure 9: Normalized Frequency Response Functions (FRFs) based on measured data (black dots) and model fits (grey line) of a representative healthy young subject for the baseline trial (0%). The GOF values indicate the goodness of fit, and are shown for each term in the FRF.
Figure 10: Estimated parameters represented by mean and standard deviation (error bars), for young and old participants per force field level. Stiffness and damping are normalized to the gravitational stiffness (mass times gravitation times CoM height) for each subject. Panel A) shows the estimated time delays, B) the intrinsic properties, C) the reflexive stiffness, and D) the reflexive damping. The asterisks represent significant differences with age (*), force field (**) or the interaction between age and force field (***).
Neuromuscular Controller

Intrinsic dynamics hip (Hp)

Receptive Field (Hr)

Intrinsic dynamics ankle (Hpa)

External Disturbances
Force2
Force1

Plant:
Double inverted pendulum

\[ \phi \]

\[ \text{Thip} \]

\[ \text{Tank} \]

Time delay

Neuromuscular Controller

\[ \phi \]

\[ \hat{\phi} \]

\[ \phi \]

\[ \hat{\phi} \]

\[ \phi \]

\[ \phi \]
Young Elderly

Ankle angle [deg]

Hip angle [deg]

2\sqrt{\lambda_1}

2\sqrt{\lambda_2}

-20 %
0 %
20 %
50 %
80 %
The graphs show the magnitude and phase responses of different joints (ank 2 Tank, hip 2 Tank, ank 2 Thip, hip 2 Thip) for young and elderly individuals across various frequencies. The magnitude is plotted on a log-scale, and the phase is plotted in degrees. Significant differences are marked with asterisks (*) and are analyzed based on the comparison between young and elderly groups.
### General characteristics

<table>
<thead>
<tr>
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<th>Young (n=15)</th>
<th>Elderly (n=14)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age, years</td>
<td>25.9 (2.8)</td>
<td>74.4 (3.5)</td>
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<tr>
<td>Women (n, %)</td>
<td>8 (53)</td>
<td>6 (43)</td>
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<tr>
<td>Weight, kg</td>
<td>71.7 (10.4)</td>
<td>78.7 (10.6)</td>
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<tr>
<td>Height, m</td>
<td>1.80 (0.09)</td>
<td>1.72 (0.08)</td>
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### Health characteristics

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<tr>
<td>Number of medication, median (IQR)</td>
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<td>2 (0-3)</td>
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<tr>
<td>MMSE, points; median (IQR)</td>
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<td>29 (28-30)</td>
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</table>

### Physical functioning

<table>
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<tr>
<td>Handgrip strength, kg</td>
<td>50.8 (16.75)</td>
<td>36.7 (7.8)</td>
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<tr>
<td>Gait speed, m/s</td>
<td>1.07 (0.16)</td>
<td>1.07 (0.15)</td>
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<tr>
<td>SPPB score, points; median (IQR)</td>
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<td>12 (11-12)</td>
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<tr>
<td>Parameter</td>
<td>Unit</td>
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<td>---------------------------------</td>
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<tr>
<td>Time delay</td>
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<tr>
<td>Ankle</td>
<td>$T_{d,\text{ank}}$</td>
<td>s</td>
</tr>
<tr>
<td>Hip</td>
<td>$T_{d,\text{hip}}$</td>
<td>s</td>
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<tr>
<td>Ankle-Hip and vice versa</td>
<td>$T_{d,\text{ank hip}}$</td>
<td>s</td>
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<td>Intrinsic feedback</td>
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<tr>
<td>Stiffness ankle</td>
<td>$K^\text{ank}_p$</td>
<td>Nm/rad</td>
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<tr>
<td>Stiffness hip</td>
<td>$K^\text{hip}_p$</td>
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<td>Reflexive feedback</td>
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<td>Stiffness Hip2Thip</td>
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<td>$D_{\text{ank} 2\text{Tank}}$</td>
<td>Nms/rad</td>
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<tr>
<td>Damping Hip2Tank</td>
<td>$D_{\text{hip} 2\text{Tank}}$</td>
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<td>Damping Ank2Thip</td>
<td>$D_{\text{ank} 2\text{Thip}}$</td>
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<tr>
<td>Damping Hip2Thip</td>
<td>$D_{\text{hip} 2\text{Thip}}$</td>
<td>Nms/rad</td>
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