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**Title**: Impaired standing balance : unraveling the underlying cause in elderly  
**Issue Date**: 2015-09-02
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Reliability of system identification techniques to assess standing balance in healthy elderly

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Abstract

System identification techniques have the potential to assess standing balance. By applying well-known disturbances, the contribution of the underlying systems in standing balance can be identified. In this study, we investigated the reliability of standing balance parameters obtained with multivariate closed loop system identification techniques.

Twelve healthy elderly participated in this study. Balance tests were performed twice a day during three days. First, body sway was tested using posturography during 2 minutes of standing with eyes closed. The Balance test Room (BalRoom) was used to apply four disturbances simultaneously: two sensory disturbances, to the proprioceptive and the visual system, and two mechanical disturbances applied at the leg and trunk segment. Using system identification techniques, sensitivity functions to the sensory disturbances and the neuromuscular controller were estimated. Systematic errors were assessed using linear mixed models, including trial and day and their interaction as fixed effects and participant intercept as random effect. Reliability was assessed using the generalizability theory, which allows for computing indexes of dependability (ID), standard error of the measurement (SEM) and minimal detectable change (MDC). To test validity, the BalRoom test was performed with increasing disturbance amplitude of the proprioceptive disturbance.

Results showed a systematic error between the first and second trial in the parameters describing the sensitivity functions. No systematic error was found in the neuromuscular controller and body sway. The reliability of the BalRoom parameters and body sway were moderate to excellent when the results of two trials on three days were averaged. To reach an excellent reliability of the BalRoom on one day, at least ten trials must be averaged. When the disturbance amplitude was increased, sensitivity functions to the proprioceptive disturbance decreased, whereas the sensitivity functions to the visual disturbances increased, which supports the validity of the method.

This study shows the possibility to use of system identification techniques to assess standing balance in elderly. As a systematic error was shown between the first and second trial on one day, it is concluded that assessment of steady state balance needs a training session and at least ten trials on one day to reach an excellent reliability.
**Introduction**

Impaired standing balance is a significant problem in elderly [1;2] and is one of the main risk factors and causes of falling [3;4]. Falls often result in serious injuries, and also in death [5]. In standing balance, several underlying systems (i.e. muscles, neural system and sensory systems) interact, which results in a closed loop system in which cause and effect are interrelated [6]. The underlying systems deteriorate with age and are influenced by diseases and medication use [7-10]. Due to redundancy, these systems can compensate for each other’s deterioration. Therefore, the underlying cause of impaired standing balance is difficult to detect and hence, to intervene with targeted therapies [11].

Current clinical balance tests do not take aforementioned cause and effect relations and redundancy of standing balance into account and therefore cannot detect the underlying cause of impaired standing balance [11]. Previous research showed that system identification techniques are useful to assess the underlying systems of standing balance, in which the response to well-known disturbances are assessed [12-16]. A clear advantage is that this method takes into account the cause and effect relation and separates the contribution of the underlying systems. This gives the opportunity to improve diagnosis of impaired balance and, eventually, to prevent falling by targeted interventions [6]. Before introducing the method into clinical practice for diagnosing or monitoring treatment of impaired balance, it is important to assess the reliability of this technique and compare it with posturography, which is yet unknown.

In this study we investigated the reliability of standing balance parameters obtained with four disturbances applied simultaneously and system identification techniques to assess standing balance in healthy elderly and compared this with a parameter obtained with posturography, namely body sway. We used the generalizability theory (G theory) [17], which takes into account both systematic and random measurement errors. Furthermore, recommendations will be given for study designs to reduce the measurement errors and therefore improve the reliability.

**Materials and methods**

**Participants**

Twelve healthy elderly aged 70 years or older participated in this study. Participants were recruited from the database of the Center of Human Drug Research, Leiden, the Netherlands,
and the MyoAge study database of the Leiden University Medical Center, Leiden, the Netherlands. Participants were screened before entry to the study. Participants were excluded in case of low cognitive function (Mini Mental State Examination (MMSE) score $\leq 26$ [18]), presence of clinical significant morbidity (haematological, renal, endocrine, pulmonary, gastrointestinal, cardiovascular, hepatic, psychiatric, neurological, musculoskeletal or allergic disorders), presence of orthostatic hypotension and use of medication. This study was approved by the Medical Ethics Committee of the Leiden University Medical Center, Leiden, the Netherlands, and was performed according to the principles of the Declaration of Helsinki and the International Conference on Harmonization/Good Clinical Practice (ICH/GCP). All participants gave written informed consent before entry to the study.

**Participant characteristics**

Prior to participation, a screening procedure was performed. Medical history was recorded including general questions about smoking, alcohol use, medication use and information on diseases. Anthropometric data included height and body composition measured with a bioelectrical impedance analysis (BIA, InBody 720, Biospace Co., Ltd, Seoul, Korea). Cognitive function was assessed with the MMSE [18]. Orthostatic hypotension was assessed by measuring blood pressure after at least 5 minutes in supine position and 3 minutes after postural change to standing position. Handgrip strength was measured using the Jamar dynamometer handle (Jamar, Sammons Preston Inc, Bolingbrook, IL, USA). Physical functioning was measured with the Short Physical Performance Battery (SPPB) [19]. Walking speed was determined by a 4 meter walking test at normal pace, as part of the SPPB.

**Apparatus**

Standing balance was assessed using the Balance test Room (BalRoom), a custom-made device applying specifically designed disturbances during stance (Motekforce Link, Culemborg, the Netherlands, and University of Twente, Enschede, the Netherlands) (Figure 1). The BalRoom consists of three separated modules. The first module consists of two support surfaces (SS), which are independently actuated and rotate around the ankles [20]. By rotation of the SS around the ankle axis the proprioceptive information of the ankle is disturbed. The second module is a visual scene (VS) in front of the participant, which rotates around the ankle axes. Rotating the VS around the ankle axis results in a disturbance of the visual information. The third module consists of two rods applying forces at hip and shoulder level (FH and FS, respectively) resulting in movements around the ankle and hip joint. These
disturbances are used to investigate the contribution of the ankles and hips and their coupling to standing balance [14]. All disturbances can be applied simultaneously.

The body sway was measured in a single plane using a string potentiometer (Celesco SP2-50, Celesco, Chatsworth, CA, United States), which integrates the amplitude of unidirectional body movement transferred through a string attached to the waist of the participant.

**Disturbance signals**

All disturbances were multisine signals with a unique combination of frequencies (Figure 2). All excited frequencies were multiples of the frequency 0.0625 Hz resulting in a disturbance period of 16 s. The SS rotated following a continuous position disturbance signal with increasing zero-to-peak amplitude over trials, i.e. 0.02, 0.03 and 0.04 radians, and a flat velocity spectrum with frequencies between 0.125 and 6.9375 Hz. The VS rotated following a continuous position disturbance signal with constant zero-to-peak amplitude of 0.03 radians over trials and a flat velocity spectrum with frequencies between 0.0625 and 1 Hz. The FH and FS disturbances are independent continuous force disturbance signals with constant zero-to-peak amplitude of 30 Newton over trials consisting of frequency contents between 0.75 and 7 Hz. All disturbances were repeated eight times resulting in a total duration of 128 seconds.

Figure 1. Set up of the Balance test Room consisting of three modules; 1) a visual scene to apply disturbances to the visual system (VS rotation), 2) support surfaces to apply disturbances to the proprioceptive system (SS rotation), and 3) two rods to apply mechanical disturbances by giving pushes and pulls at hip and shoulder level (FH and FS).
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Figure 2. Normalized time signals and frequency spectra of the disturbances of the support surface (SS) rotation, the visual scene (VS) rotation and the rods applying forces at hip and shoulder level (FH and FS, respectively).

Procedure

During the screening visit for inclusion up to 21 days before the start of the study, each participant had a training session to familiarize with the BalRoom test and disturbances and with the body sway test. No data were recorded. During the study, the tests were performed during three sessions separated by one week, allowing assessment of intersession variability. Per session the tests were performed twice separated by one hour, allowing assessment of intrasession variability. During all tests the participant wore comfortable flat shoes. During the BalRoom test, the participant was instructed to stand with the arms resting along the body, with both feet in place on the support surfaces. The two sensory (SS and VS) and mechanical (FH and FS) disturbances were applied simultaneously. Each test consisted of three conditions with increasing disturbance amplitude of the SS rotation (i.e. 0.02, 0.03 and 0.04 radians), while the amplitudes of the VS, FH and FS disturbances remained constant. The three conditions were presented in random order. Before recording each condition the participant was allowed about 10 seconds to get accustomed to the disturbances. Between conditions, the
participant was offered ample resting time depending on individual needs. The participant wore a safety harness to prevent falling, which did not constrained movement nor provided support or orientation information.

During the body sway test, the participant was asked to stand still and comfortable for a period of 2 minutes, with the feet approximately 10 cm apart and the hands in a relaxed position along the body and eyes closed.

**Data recording and processing**

The actual angles of SS rotation (i.e. motor angles), applied forces at hip and shoulder level (FH and FS forces) and the applied torques to the SS (i.e. motor torques) were available for measurement. Lower and upper body segmental movements were measured in anterior-posterior direction using two draw wire potentiometers (Celesco SP2-50, Celesco, Chatsworth, CA, United States) at a sample frequency of 1000 Hz. The potentiometers were connected to the hip and the shoulders by magnets and straps. The motor angles, segment angles,motor torques and applied FH and FS forces were recorded using a Matlab interface with a sample frequency of 1000 Hz. Data analysis was performed with Matlab (The MathWorks, Natick, MA, United States). The leg and hip angle were calculated using goniometry and using the segment movement of the lower and upper body [21]. The ankle torque was obtained by subtracting the contribution of the mass and inertia of the support surfaces from the recorded motor torques. The hip torque was obtained using the applied FH and FS forces and leg and hip angles using inverse dynamics [21]. The time series were segmented into eight data blocks of 16 seconds (i.e. the period of the disturbance signal).

**Data analysis**

To indicate the effect of the disturbances on the ankle torque, hip torque and joint angles, Frequency Response Functions (FRFs) were estimated. The time series of the disturbances, ankle torque, hip torque, leg and hip angle were transformed to the frequency domain. The periodic part of the frequency coefficients was determined by averaging over the data blocks. The Power Spectral Densities (PSD) and Cross Spectral Densities (CSD) were computed to calculate the FRFs [22]. For each disturbance, only the excited frequencies were analyzed.
Sensitivity functions

The sensitivity functions of the ankle torque, hip torque, leg angle and hip angle to the SS rotation and the VS rotation were estimated using the indirect approach using equation 1 [12;22].

\[ dS_x(f) = \Phi_{d,x}(f) \cdot [\Phi_{d,d}(f)]^{-1} \]  

\( (1) \)

In which \( \Phi_{d,x} \) represents the CSD of the disturbance (d) (i.e. SS rotation or VS rotation) and x, which represents the ankle torque (T_a), hip torque (T_h), leg angle (\( \theta_l \)), or hip angle (\( \theta_h \)), and \( \Phi_{d,d} \) the PSD of the disturbance. This results in 8 FRFs; 1) SS rotation to ankle torque (\( ^{SS}S_{Ta} \)), 2) SS rotation to hip torque (\( ^{SS}S_{Th} \)), 3) SS rotation to leg angle (\( ^{SS}S_{\theta_l} \)), and 4) SS rotation to hip angle (\( ^{SS}S_{\theta_h} \)), and 5) to 8) the VS rotation to each torque and angle (\( ^{VS}S_{Ta} \), \( ^{VS}S_{Th} \), \( ^{VS}S_{\theta_l} \), \( ^{VS}S_{\theta_h} \)). Each FRF is represented by a magnitude and phase representing the ratio between the input and output and the relative timing both as function of frequency. The magnitude of the sensitivity function of the ankle and hip torque is normalized to the gravitational stiffness (mg\( l_{CoM} \)). The average magnitude on the low frequencies (<0.375 Hz and <0.1875 Hz, for SS and VS respectively) and the phase on higher frequencies (0.68 Hz and 0.375 Hz, for SS and VS respectively) are the parameters of interest. Different values of frequencies were used for SS and VS due to differences in frequency content. They represent the sensitivity to the disturbances and the time lag between the disturbance and the reaction of the body, respectively, resulting in 16 parameters.

Neuromuscular controller

The neuromuscular controller is the link between the sensory systems and the muscles, where the sensory information is combined and muscle commands are generated to keep the body in upright position. The multi-input-multi-output (MIMO) approach was used to estimate the ankle and hip controller according to the method described by Engelhart et al. (2014) and equation 2 [14].

\[ H_c(f) = -\Phi_{d,T}(f) \cdot [\Phi_{d,\theta}(f)]^{-1} \]  

\( (2) \)

In which \( \Phi_{d,T} \) and \( \Phi_{d,\theta} \) are the CSD matrices between the external disturbance (d) (i.e. FH and FS) and the corrective ankle and hip torques (T) and the leg and hip angles (\( \theta \)) resulting in a two-by-two FRF matrix (Hc). This results in 4 FRFs; 1) ankle torque to ankle angle (\( H_{Ta2\theta} \)), 2) ankle torque to hip angle (\( H_{Ta2\theta_h} \)), 3) hip torque to hip angle (\( H_{Th2\theta_h} \)), and 4) hip torque to ankle angle (\( H_{Th2\theta_l} \)). The magnitude is normalized to the gravitational stiffness (mg\( l_{CoM} \)). The average magnitude on the low frequencies (<1 Hz) and the phase on higher frequencies (2.3
Hz) are the parameters of interest and represent the normalized effective stiffness and the time lag between the torques and angles, resulting in 8 parameters [14].

Body sway

The body sway ($x_{BS}$) was measured over 2 minutes. The movement of the body was expressed as millimeters of sway.

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. The parameters obtained with system identification techniques (i.e. sensitivity and time lag of the sensitivity functions, and normalized effective stiffness and time lag of the neuromuscular controller) are given as mean and standard deviation.

Reliability was assessed using the G theory in three steps [17]. First, systematic errors were identified using linear mixed models with trial (intrasession), day (intersession) and their interaction as fixed effects and participant intercept as random effect. The various sources of measurement errors were assessed using a random effects repeated measures analysis of variance (ANOVA) including participant, trial, day and their interactions. This resulted in the variance of the participants ($\sigma^2_p$), the variance of the trials ($\sigma^2_t$), the variance of the day ($\sigma^2_d$), the variance of their interactions ($\sigma^2_{pt}$, $\sigma^2_{pd}$ and $\sigma^2_{td}$) and the variance of the residual ($\sigma^2_{ptd,e}$). All were presented as percentages of the total variance. Negative variance components were set to zero. The sources of variance were used to calculate the index of dependability (ID), the standard error of the measurement (SEM) and the minimal detectable change (MDC) using equation 3 [17,23].

$$\sigma^2_\Delta = \frac{\sigma^2_i}{n_i} + \frac{\sigma^2_d}{n_d} + \frac{\sigma^2_{pt}}{n_i n_d} + \frac{\sigma^2_{pd}}{n_i n_d} + \frac{\sigma^2_{td}}{n_i n_d} + \frac{\sigma^2_{ptd,e}}{n_i n_d}$$

$$ID = \frac{\sigma^2_p}{\sigma^2_\Delta}$$

$$SEM = \sqrt{\sigma^2_\Delta}$$

$$MDC = 1.96 \times \sqrt{2} \times SEM$$

In which, nt is the number of trials and nd the number of days.

Comparable with an intraclass correlation coefficient (ICC), the ID ranges between 0 and 1 and can be interpreted as; ID < 0.40 represents poor reliability, 0.40 < ID < 0.75 represents
moderate reliability, and ID > 0.75 represents excellent reliability [24]. The SEM indicates the absolute reliability and is represented by an absolute value and a percentage of the overall mean. The MDC shows which effect (e.g. treatment effect) can be detected with the parameters of interest and therefore indicates the clinical relevance. A low SEM and MDC are indicative of a reliable and clinical relevant parameter.

Second, a decision study was performed in which the effect of different measurement protocols on the reliability was investigated. Aforementioned equations show that increasing the number of trials or number of days results in an increase of ID and a decrease of SEM and MDC, i.e. an improvement of reliability. In the decision study, the number of trials was varied between 1 and 40 trials and the number of days between 1 and 3. Per number of days, the number of trials needed to reach an excellent reliability was determined in this group of healthy elderly (ID > 0.75).

Third, a validity study was performed to assess whether differences in the sensitivity functions represented by the sensitivity and time lag due to increasing disturbance amplitude of the SS rotation could be detected. A linear mixed model was constructed with disturbance amplitude as fixed effect and participant intercept as random effect. Statistical analysis was performed with SPSS version 20 (SPSS Inc., Chicago, USA). Graphs were made with Matlab (The MathWorks, Natick, MA, United States).

Results

Participant characteristics

Table 1 presents the characteristics of the healthy old participants. Figure 3, 4 and 5 displays the FRFs of the sensitivity functions to the SS rotations and the VS rotations, and of the neuromuscular controller.

Systematic errors

Table 2 reports the systematic errors. No systematic errors were found for the body sway ($x_{BS}$). The sensitivity functions show both a main effect of trial and day. Overall, the sensitivity to the SS rotation was lower during the first trial compared with the second trial in the sensitivity functions (i.e. $SS_{Th}$, $SS_{q}$ and $SS_{Ta}$) and it was lower during the first day compared to the second and third day (for $SS_{Th}$ and $SS_{Ta}$).
The sensitivity function to the VS rotation shows the opposite result; the sensitivity of the first trial was higher compared with the second trial for all sensitivity functions ($V_S^{S_{0b}}$, $V_S^{S_{0h}}$, $V_S^{S_{Th}}$ and $V_S^{S_{Ta}}$). Furthermore, the time lags of some sensitivity functions (i.e. $S_{Ta}^{S_{Ta}}$ and $V_S^{S_{0h}}$) were higher in the first trial compared with the second trial. The time lags of $V_S^{S_{Ta}}$ and $V_S^{S_{Th}}$ were also lower during the first day compared with the third day.

The normalized effective stiffness estimated using the FS and FH disturbances showed an effect of the trial; one component of the neuromuscular controller ($H_{0|Ta}$) was lower during the first trial compared with the second trial. No effect of trial and day was found for the time lags of all components of the neuromuscular controller.

**Variance components**

Table 3 shows the magnitude of the variance components as percentage of the total variance. Variance of the participant ($\sigma_p^2$) was on average 20.8% and varied between 0.0% and 87.3%. The contribution of the trial variance ($\sigma_t^2$) was on average 6.6% and varied between 0.0% and 28.8%. The contribution of the day variance ($\sigma_d^2$) was on average 2.5% and varied between 0.0% and 12.5%.

### Table 1. Participant characteristics.

|                                | All (n=12)     |
|--------------------------------|----------------|
| Age, years                     | 73.3 (3.4)     |
| Men, n (%)                     | 6 (50)         |
| **Anthropometry**              |                |
| Weight, kg                     | 72.2 (9.1)     |
| Height, m                      | 1.70 (0.08)    |
| BMI, kg/m$^2$                  | 24.9 (2.4)     |
| **Health characteristics**     |                |
| Number of medication, median (IQR) | 0 (0-0)       |
| MMSE, points; median (IQR)     | 29 (28-30)     |
| **Physical function**          |                |
| Handgrip strength, kg          | 34.7 (8.6)     |
| Gait speed, m/s                | 1.28 (0.16)    |
| SPPB, points; median (IQR)     | 11 (10-12)     |

All parameters are presented as mean with standard deviation unless indicated otherwise. BMI: body mass index, MMSE: Mini Mental State Examination, SPPB: Short Physical Performance Battery, IQR: inter quartile range.
Figure 3. Average sensitivity functions of the ankle torque ($SS_{Ta}$), hip torque ($SS_{Th}$), leg angle ($SS_{θl}$) and hip angle ($SS_{θh}$) to the rotation of the support surfaces per day per trial are presented by mean and standard error, only magnitude is shown.

The error variance related to the interactions between the participant and trial ($σ_{pt}^2$), between participant and day ($σ_{pd}^2$) and between trial and day ($σ_{td}^2$) were low; on average they were 4.3%, 13.4% and 4.4%, respectively.

The largest proportion of measurement variability was due to the participant variability ($σ_p^2$) and the other interactions combined with the residual error ($σ_{ptd,e}^2$) contributing on average 48.0% ranging from 7.2% to 80.8%.
Figure 4. Average sensitivity functions of the ankle torque ($v_{SA}$), hip torque ($v_{SB}$), leg angle ($v_{S\theta L}$) and hip angle ($v_{S\theta H}$) to the rotation of the visual scene per day per trial are presented by mean and standard error, only magnitude is shown.

Reliability

Table 4 presents the results of the reliability measures. In this study design, the ID of 4 out of 25 parameters was higher than 0.75 and in 12 out of 25 parameters ID was between 0.40 and 0.75. The SEM and SEM % were inverse related with the ID. Furthermore, the MDC was lower with increased ID. To reach an ID of 0.75, for 32% (8/25) of the parameters at least ten trials are needed to average over one day. Increasing the number of days resulted in less trials needed per day to reach an ID higher than 0.75.
Figure 5. Average Frequency Response Functions of the neuromuscular controller (i.e. $H_{T_a2\theta}$, $H_{T_2a2\theta}$, $H_{T_2\theta}$, $H_{T_2\theta}$) per day per trial are presented by mean and standard error, only magnitude is shown.

Validity

Table 5 presents the results of the validity study. The mean and standard deviation of the parameters of the second trial at the first day are given for each condition. All sensitivities to the SS rotation decreased with increasing disturbance amplitude and all sensitivities to the VS rotation increased with increasing disturbance amplitude. No significant differences were found for the time lag of the sensitivities to the SS rotation and VS rotation. No significant differences were found between the conditions for the parameters describing the neuromuscular controller.
Discussion

In this study, we assessed the reliability of a comprehensive set of parameters obtained with four disturbances applied simultaneously and (MIMO closed loop) system identification techniques describing standing balance in a group of healthy elderly. Results were obtained by measuring standing balance twice during three days. A distinction was made between systematic and random errors. The results showed a systematic error between the first and second trial with the BalRoom on one day, which was not found using body sway measurements. The reliability ranged from moderate to excellent when averaging two trials on three days. This is the first study that investigated the reliability of system identification techniques to assess standing balance in healthy elderly.

Systematic errors

In general, the sensitivity to the SS rotation was lower in the first trial compared with the second trial, while the sensitivity to the VS rotation was higher in the first trial compared to the second trial. These results are confirmed by the variance component of the trial ($\sigma_t^2$) and day ($\sigma_d^2$); a high variance component of trial and day indicates a systematic error. Previous studies using system identification techniques also showed a systematic error between the first and second trial or between days. These differences were explained by motor learning, changes in posture or stretching of the joints [25-27]. In contrast, in a previous study no learning effects were found. These results might be due to the practice session all participants performed prior to participation in this study [28].

In this study, the differences between the first trial compared with the second trial (i.e. a lower sensitivity to proprioception and a higher sensitivity to vision during the first trial) could be explained by a difference in strategy used to maintain standing balance or familiarization during the test.

According to the sensory reweighting hypothesis, sensory information is weighted based on reliability; the weight of the proprioception increased at the cost of a decrease of the weight of the other sensory information [12]. As the sensitivity to the disturbances represents the contribution of the proprioceptive and visual information, the sensitivity to the SS rotation increases, while the sensitivity to the VS rotation decreases.
Table 2. Systematic errors of all parameters using linear mixed model with day, trial and their interaction as fixed effect and subject intercept as random effect.

|                      | Mean   | SD     | p-value | Post hoc analyse |
|----------------------|--------|--------|---------|-----------------|
|                      | trial (t) | day (d) | t x d   | trial | day |
| **Body sway**        |         |        |         |       |     |
| $x_{BS}$, mm         | 330.53  | 139.12 | 0.18 0.47 0.29 | - | -   |
| **Sensitivity functions** |   |         |         |       |     |
| Sensitivity          |         |        |         |       |     |
| $SS_{S_{th}}$        | 1.05   | 0.38   | **0.005** 0.19 0.008 | Day 1 and 2: |             |
|                      |        |        |         | Trial 1 < Trial 2 | - |
| $SS_{S_{th}}$        | 0.65   | 0.17   | <0.001 0.05 0.076 | Trial 1 < Trial 2 | - |
| $SS_{S_{Th}}$        | 0.99   | 0.27   | **0.001** 0.007 0.38 | Trial 1 < Trial 2 | Day 1 < Day 3 |
|                      |        |        |         |       |     |
| **Time lag**         |         |        |         |       |     |
| $SS_{S_{th}}$, deg   | 118.42 | 22.60  | 0.16 0.68 0.30 | - | -   |
|                      |        |        |         |       |     |
| **Neuromuscular controller** | | |         |       |     |
| Normalized effective stiffness |         |        |         |       |     |
| $H_{th\theta_{th}}$  | 0.70   | 0.33   | 0.93 0.83 0.96 | - | -   |
|                      | 0.21   | 0.10   | 0.085 0.022 0.12 | - | Day 1 > Day 2 |
| $H_{th\theta_{Ta}}$  | 3.70   | 1.30   | 0.83 0.21 0.86 | - | -   |
| $H_{th\theta_{Th}}$  | 0.36   | 0.21   | **0.015** 0.75 0.85 | Trial 1 < Trial 2 | - |
| **Time lag**         |         |        |         |       |     |
| $H_{th\theta_{th}}$, deg | -109.72 | 61.91 | 0.72 0.21 0.25 | - | -   |
|                      | 96.07  | 31.93  | 0.42 0.10 0.73 | - | -   |
| $H_{th\theta_{Th}}$, deg | -112.58 | 38.74 | 0.83 0.47 0.80 | - | -   |
| $H_{th\theta_{Ta}}$, deg | 117.88 | 28.01 | 0.48 0.34 0.86 | - | -   |

Significant differences identified in bold. n.s. : not significant.
Table 3. Relative magnitude of the variance components obtained with the G-study for all parameters obtained with system identification techniques.

|                         | participant (p), % | trial (t), % | day (d), % | p x t, % | p x d, % | t x d, % | p x t x d, e, % |
|-------------------------|--------------------|--------------|------------|----------|----------|----------|-----------------|
| **Body sway**           |                    |              |            |          |          |          |                 |
| $x_{RS}$                | 87.3               | 0.2          | 0.0        | 0.0      | 4.7      | 0.5      | 7.2             |
| **Sensitivity functions**|                    |              |            |          |          |          |                 |
| Sensitivity             |                    |              |            |          |          |          |                 |
| $SS_{S \theta h}$       | 24.9               | 5.5          | 0.0        | 3.1      | 8.7      | 20.7     | 37.1            |
| $SS_{S \theta l}$       | 11.0               | 28.8         | 0.0        | 4.7      | 9.4      | 9.8      | 36.5            |
| $SS_{S \tau a}$         | 19.6               | 15.5         | 6.7        | 0.0      | 11.0     | 2.2      | 44.9            |
| $SS_{S \tau b}$         | 11.2               | 0.0          | 12.5       | 4.0      | 23.2     | 16.9     | 32.1            |
| $VS_{S \theta h}$       | 26.0               | 10.1         | 0.3        | 0.0      | 0.0      | 0.0      | 63.6            |
| $VS_{S \theta l}$       | 0.0                | 16.2         | 0.0        | 22.2     | 6.7      | 0.0      | 54.9            |
| $VS_{S \tau a}$         | 0.0                | 23.7         | 3.5        | 19.3     | 17.2     | 0.0      | 36.2            |
| $VS_{S \tau b}$         | 39.1               | 16.7         | 0.0        | 0.0      | 5.9      | 3.6      | 34.8            |
| **Time lag**            |                    |              |            |          |          |          |                 |
| $SS_{S \theta h}$       | 27.7               | 1.6          | 0.0        | 1.4      | 0.0      | 6.3      | 63.0            |
| $SS_{S \theta l}$       | 6.9                | 6.4          | 0.0        | 0.0      | 0.0      | 5.9      | 80.8            |
| $SS_{S \tau a}$         | 18.9               | 0.0          | 0.0        | 6.0      | 12.8     | 17.2     | 45.1            |
| $SS_{S \tau b}$         | 29.6               | 0.0          | 0.0        | 0.0      | 9.4      | 6.0      | 55.0            |
| $VS_{S \theta h}$       | 7.5                | 5.4          | 6.0        | 13.5     | 14.3     | 0.0      | 53.3            |
| $VS_{S \theta l}$       | 3.9                | 16.0         | 0.0        | 12.7     | 21.3     | 4.2      | 41.9            |
| $VS_{S \tau a}$         | 41.8               | 0.0          | 2.3        | 0.0      | 0.0      | 5.1      | 50.9            |
| $VS_{S \tau b}$         | 20.3               | 0.0          | 10.0       | 0.2      | 11.9     | 0.8      | 56.8            |
| **Neuromuscular controller**|                    |              |            |          |          |          |                 |
| Normalized effective stiffness |                |              |            |          |          |          |                 |
| $H_{\theta 2Ta}$        | 48.1               | 1.2          | 1.8        | 0.0      | 0.0      | 0.0      | 48.9            |
| $H_{\theta 2Th}$        | 34.6               | 1.7          | 2.5        | 0.0      | 4.7      | 5.0      | 51.5            |
| $H_{\tau 2Ta}$          | 48.9               | 1.0          | 3.4        | 0.0      | 0.0      | 0.0      | 46.8            |
| $H_{\tau 2Th}$          | 21.7               | 9.0          | 2.4        | 0.0      | 0.0      | 0.0      | 66.9            |
| **Time lag**            |                    |              |            |          |          |          |                 |
| $H_{\theta 2Ta}$        | 18.4               | 0.0          | 0.0        | 0.0      | 33.5     | 0.3      | 47.8            |
| $H_{\theta 2Th}$        | 7.8                | 0.0          | 6.0        | 7.9      | 55.3     | 1.2      | 21.7            |
| $H_{\tau 2Ta}$          | 19.0               | 0.0          | 0.0        | 4.9      | 25.2     | 0.0      | 50.9            |
| $H_{\tau 2Th}$          | 13.4               | 0.0          | 1.4        | 4.2      | 50.1     | 0.5      | 30.3            |
| **Mean**                | **20.8**           | **6.6**      | **2.5**    | **4.3**  | **13.4** | **4.4**  | **48.0**        |
The combination of mechanical disturbances with sensory disturbances of the visual and proprioceptive information could have resulted in a longer adaptation time or a redundancy of applied strategies to withstand the disturbances. However, comparable systematic errors within a day were found in healthy elderly (unpublished data) in a previous study using only SS rotation to disturb proprioceptive information (Pasma et al. submitted), which suggest that the longer adaptation time is not due to the combination of multiple disturbances. In contrast, no systematic errors were found in healthy young adults (unpublished data). This is an indication of increased adaptation time in elderly compared with young adults.

When a steady state of standing balance is to be assessed, a familiarization trial is needed on the same day to overcome the systematic error between trials. Excluding the first trial of each day resulted in less systematic errors between days.

**Reliability**

The variance component of the participant ($\sigma_p^2$) corresponds to the ICC when both nt and nd are equal to one. The reliability of the parameters ranged from poor to moderate. To increase reliability of steady state balance assessment, multiple trials on more than one day have to be performed. The ID values indicate that performing two trials on three days results in a reliability ranging from moderate to excellent needed to discriminate between healthy old individuals. A high residual variance ($\sigma_{ptd,e}^2$) component indicates that a majority of the measurement error is random or can be attributed to error sources not identified in the study.

In this study, relative low SEM% were found (<20%), which is comparable with other studies using system identification techniques [25]. A low SEM% indicates that the parameter could detect changes over time within the same participant (e.g. effects of intervention or changes in conditions). However, the SEM values depend on the number of trials performed on the number of days. The MDC values are in the same order as in a previous study using only SS rotation in healthy elderly (unpublished data) and indicates which change in the parameters can be minimally detected, when comparing groups or within the same participant. It is difficult to interpret the MDC results of new parameters. To get more feeling for this measure and to get more insight in the clinical relevance, it is recommended to assess standing balance using system identification techniques in several groups of elderly with a large variance in impaired balance severity and clinical phenotypes [23].
Table 4. Reliability statistics.

| ID  | SEM  | SEM % | MDC | # trials / 1 day | # trials / 2 days | # trials / 3 days |
|-----|------|-------|-----|------------------|-------------------|------------------|
|     |      |       |     | >0.75            | >0.75             | >0.75            |

**Body sway**

| xBS | 0.97 | 25.01 | 69.32 | 1                | 1                 | 1                |

**Sensitivity functions**

|     |      |       |     |                  |                   |                   |
|-----|------|-------|-----|------------------|-------------------|------------------|
|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |

|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |

**Time lag**

|     |      |       |     |                  |                   |                   |
|-----|------|-------|-----|------------------|-------------------|------------------|
|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |
|     |      |       |     |                  | >40               |                   |

**Neuromuscular controller**

|     |      |       |     |                  |                   |                   |
|-----|------|-------|-----|------------------|-------------------|------------------|
|     |      |       |     |                  |                   |                   |
|     |      |       |     |                  |                   |                   |
|     |      |       |     |                  |                   |                   |

ID = Index of Dependability (>0.75 identified in bold), SEM = Standard Error of Measurement, MDC = Minimal Detectable Change
The results showed that at least 10 trials on one day are needed to reach an excellent reliability for steady state balance assessment in one third of the parameters. In this study, averaging trials across days seems to be more effective than averaging more trials per day. These results are consistent with the variance component of interaction; the variance component of participant x day ($\sigma_{pd}^2$) is much higher than the variance component of participant x trial ($\sigma_{pt}^2$). This means that the parameters for each participant were more affected by between day than within day sources of error, relative to the other participants. These results are in accordance with previous studies; Lariviere et al. (2015) also showed that one till ten trials were needed to assess an excellent reliability for parameters obtained with system identification techniques [25]. A lower reliability seems to be a general feature of position stabilization task in contrast to tracking tasks [26].

**Validity**

The validity study showed that differences could be detected within participants by changing the experimental condition. It was possible to detect changes over conditions using one trial. Increasing the disturbance amplitude of the SS rotation resulted in a decreased sensitivity to the SS rotation and an increased sensitivity to the VS rotation. This result was expected according to the sensory reweighting hypothesis, as mentioned before. Our findings are therefore also in line with previous studies investigating sensory reweighting during standing balance using system identification techniques [12;15].

No changes were found in the neuromuscular controller by increasing the disturbance amplitude of the SS rotation. This is following our expectations, as changes in sensory information does not influence the stiffness and damping of the neuromuscular controller. These results are also in accordance with a previous study, in which we showed that the neuromuscular controller did not change with increasing disturbance amplitude of the SS rotation [15].

**System identification techniques compared to posturography**

System identification techniques are a new engineering approach to assess standing balance. In contrast with posturography, a general used technique to assess standing balance, it is possible to detect underlying systems and used strategies in standing balance [6;11]. In this study, we assessed standing balance with both system identification techniques and posturography (i.e. body sway). Compared to system identification techniques, no systematic errors and a higher reliability were found for posturography.
Table 5. Mean and standard deviation of the parameters corresponding to three conditions with increasing disturbance amplitude, combined with statistical results.

|                  | 0.02 rad | 0.03 rad | 0.04 rad | p-value |
|------------------|----------|----------|----------|---------|
|                  | Mean     | SD       | Mean     | SD      | Mean  | SD      |            |         |
| **Sensitivity functions** |          |          |          |         |       |         |            |         |
| Sensitivity      |          |          |          |         |       |         |            |         |
| $SS_{\theta h}$  | 1.13     | 0.31     | 0.94     | 0.41    | 0.82  | 0.24    | **0.003** |         |
| $SS_{\theta l}$  | 0.73     | 0.14     | 0.58     | 0.13    | 0.48  | 0.09    | **<0.001** |         |
| $SS_{\theta h}$  | 1.08     | 0.21     | 0.93     | 0.20    | 0.75  | 0.13    | **<0.001** |         |
| $SS_{\theta l}$  | 0.19     | 0.05     | 0.18     | 0.05    | 0.15  | 0.03    | **0.001** |         |
| $SS_{Ta}$        | 0.73     | 0.22     | 0.82     | 0.28    | 0.93  | 0.27    | 0.18       |         |
| $SS_{Th}$        | 0.58     | 0.14     | 0.63     | 0.16    | 0.77  | 0.28    | 0.079      |         |
| $VS_{\theta h}$  | 0.73     | 0.25     | 0.90     | 0.23    | 1.04  | 0.39    | **0.044**  |         |
| $VS_{\theta l}$  | 0.12     | 0.04     | 0.14     | 0.05    | 0.17  | 0.07    | **0.015**  |         |
| **Time lag**     |          |          |          |         |       |         |            |         |
| $SS_{\theta h}$  | 114.74   | 19.76    | 114.47   | 21.32   | 115.66| 19.37   | 0.98       |         |
| $SS_{\theta l}$  | -98.15   | 24.46    | -105.43  | 20.26   | -92.20| 30.52   | 0.42       |         |
| $SS_{Ta}$        | 16.34    | 11.98    | 15.39    | 8.90    | 22.39 | 12.54   | 0.094      |         |
| $SS_{Th}$        | -37.66   | 32.00    | -41.71   | 12.53   | -41.50| 15.35   | 0.81       |         |
| $VS_{\theta h}$  | 52.80    | 117.92   | 23.45    | 140.10  | 94.19 | 42.41   | 0.32       |         |
| $VS_{\theta l}$  | -96.04   | 31.73    | -105.29  | 44.74   | -102.27| 19.63   | 0.80       |         |
| $VS_{Ta}$        | 43.44    | 25.46    | 36.72    | 27.95   | 40.84 | 18.80   | 0.73       |         |
| $VS_{Th}$        | -0.46    | 32.38    | -9.24    | 31.00   | -9.29 | 24.15   | 0.33       |         |
| **Neuromuscular controller** |          |          |          |         |       |         |            |         |
| Normalized effective stiffness |          |          |          |         |       |         |            |         |
| $H_{\theta hTa}$ | 0.70     | 0.26     | 0.80     | 0.49    | 0.75  | 0.28    | 0.61       |         |
| $H_{\theta lTa}$ | 0.18     | 0.05     | 0.20     | 0.08    | 0.19  | 0.11    | 0.74       |         |
| $H_{Ta}$         | 3.70     | 1.21     | 3.93     | 1.60    | 3.93  | 1.03    | 0.80       |         |
| $H_{Th}$         | 0.42     | 0.25     | 0.39     | 0.22    | 0.29  | 0.15    | 0.29       |         |
| **Time lag**     |          |          |          |         |       |         |            |         |
| $H_{\theta hTa}$ | -89.99   | 73.93    | -76.50   | 66.69   | -78.23| 78.66   | 0.89       |         |
| $H_{\theta lTa}$ | 102.46   | 39.66    | 119.09   | 44.69   | 119.24| 36.63   | 0.42       |         |
| $H_{Ta}$         | -110.27  | 58.82    | -86.77   | 37.46   | -108.93| 48.49   | 0.28       |         |
| $H_{Th}$         | 123.08   | 36.14    | 136.45   | 31.24   | 138.22| 29.81   | 0.37       |         |

In comparison with our results of system identification techniques, studies investigating the reliability of the Sensory Organization Test (SOT) showed a learning effect in healthy young due to changes in postural strategies or through reweighting of sensory information. Remarkably, this learning effect was only present in more demanding test conditions [29]. Studies investigating the reliability of Center of Pressure (CoP) parameters did not find
systematic errors [30;31], which is comparable with our results of the body sway, but in contrast with the system identification techniques results. This could be explained by the influence of used strategies to maintain balance on the parameters. CoP parameters only describe objectively standing balance, while system identification techniques also describe the underlying changes. Therefore, changes in strategies between trials will not be detected by CoP parameters and do not influence the reliability of CoP parameters.

The reliability of the SOT was moderate in noninstitutionalized old adults when 2 sessions of the test were performed 1 week apart. To improve the reliability of the computer-generated scores of the SOT, a modification of the scoring system was recommended [32]. The reliability of CoP parameters depends on the test condition, study design, study population and therapeutic interventions [33]. To reach an excellent reliability of CoP parameters, the duration of the trial must be minimal 90 seconds, must by three to five times repeated and must be measured with eyes closed and on a firm surface [33]. Santos et al. (2007) showed that at least 10 repetitions must be performed to reach an excellent reliability for CoP parameters [34]. This is comparable with our study, in which measurements of approximately two minutes were used to assess standing balance and must be repeated ten times to reach an excellent reliability. The found relative low SEM% (<20%) are comparable with other studies using CoP parameters [34].

**Clinical recommendations**

First, the results indicated that there is a systematic error between the first trial and the second trial. This could be due to changes in used strategies to maintain standing balance and time needed to reach a steady state. Therefore, to assess steady state balance we recommend to perform one familiarization trial on each day. As in this study only two trials were performed per day, it was not possible to assess the number of trials needed to reach an excellent reliability when omitting the first trial from analysis. Second, results showed that averaging over days is more effective than averaging within days. However, in clinical practice it is not feasible to measure on more than one day. Performing multiple measurements on one day could be hampered by fatigue or boredom of the participant, which has to be taken into account.

As mentioned before, systematic errors might be due to more time needed for reaching a steady state balance or a redundancy of applied strategies. This implies that parameters obtained with system identification techniques are sensitive for detection of adaptation
strategies. Besides steady state balance, adaptation strategy and adaptation time may have clinical meaning and need further exploration. System identification techniques are sensitive tools to assess the duration of adaptation of sensory reweighting [35] in contrast to e.g. CoP measurement.

**Strengths and limitations**

The strength of this study is the selection of healthy old participants, resulting in a well phenotyped group. However, this also affects ICC and ID. Low variability within the participants (i.e. a homogeneous population) results in lower ICC and ID values and therefore lower relative reliabilities [36;37]. SEM(%) and MDC are measures of absolute reliability and important measures when interpreting results of repeated measures effects of intervention. Another strength of this study is the set up with exactly one week between sessions. A limitation of this study is the relative low number of participants. As only two trials per day were performed, reliability within a day could not be assessed excluding the first (familiarization) trial.

**Conclusion**

This study investigated the reliability of a comprehensive set of parameters obtained with system identification techniques to assess standing balance in a population of healthy elderly. Systematic errors were present between trials showing sensitivity of parameters obtained with system identification techniques for detection of adaptation strategies. To assess steady state balance a training session is recommended. As only a single trial per day resulted in poor to moderate reliability, it is recommended to perform more trials on separate days. Within the present framework, acceptable reliability of steady state balance assessment could be achieved by measuring and averaging at least ten trials on the same day.

**Acknowledgements**

This research (BalRoom; project number 10737) is supported by the Dutch Technology Foundation STW, which is part of the Netherlands Organization for Scientific Research (NWO) and which is partly funded by the Ministry of Economic Affairs.
References

1. Jonsson R, Sixt E, Landahl S, Rosenhall U. Prevalence of dizziness and vertigo in an urban elderly population. J Vestib Res 2004;14(1):47-52.

2. Lin HW, Bhattacharyya N. Balance disorders in the elderly: epidemiology and functional impact. Laryngoscope 2012;122(8):1858-61.

3. Rubenstein LZ. Falls in older people: epidemiology, risk factors and strategies for prevention. Age Ageing 2006;35 Suppl 2:i37-i41.;i37-i41.

4. Muir SW, Berg K, Chesworth B, Klar N, Speechley M. Quantifying the magnitude of risk for balance impairment on falls in community-dwelling older adults: a systematic review and meta-analysis. J Clin Epidemiol 2010;63(4):389-406.

5. Cummings SR, Kelsey JL, Nevitt MC, O'Dowd KJ. Epidemiology of osteoporosis and osteoporotic fractures. Epidemiol Rev 1985;7:178-208.

6. Engelhart D, Pasma JH, Schouten AC, Meskers CG, Maier AB, Mergner T, et al. Impaired standing balance in elderly: a new engineering method helps to unravel causes and effects. J Am Med Dir Assoc 2014;15(3):227-6.

7. Horak FB, Shupert CL, Mirka A. Components of postural dyscontrol in the elderly: a review. Neurobiol Aging 1989;10(6):727-38.

8. Manchester D, Woollacott M, Zederbauer-Hylton N, Marin O. Visual, vestibular and somatosensory contributions to balance control in the older adult. J Gerontol 1989;44(4):M118-M127.

9. Konrad HR, Girardi M, Helfert R. Balance and aging. Laryngoscope 1999;109(9):1454-60.

10. Sturnieks DL, St GR, Lord SR. Balance disorders in the elderly. Neurophysiol Clin 2008;38(6):467-78.

11. Pasma JH, Engelhart D, Schouten AC, van der Kooij H, Maier AB, Meskers CG. Impaired standing balance: the clinical need for closing the loop. Neuroscience 2014;267:157-65:157-65.

12. Peterka RJ. Sensorimotor integration in human postural control. J Neurophysiol 2002;88(3):1097-118.

13. Cenciarini M, Loughlin PJ, Sparto PJ, Redfern MS. Stiffness and damping in postural control increase with age. IEEE Trans Biomed Eng 2010;57(2):267-75.

14. Engelhart D, Schouten AC, Aarts RG, van der Kooij H. Assessment of Multi-Joint Coordination and Adaptation in Standing Balance: a Novel Device and System Identification Technique. IEEE Trans Neural Syst Rehabil Eng 2014;Epub ahead of print.
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15. Pasma JH, Boonstra TA, Campfens SF, Schouten AC, van der Kooij H. Sensory reweighting of proprioceptive information of the left and right leg during human balance control. J Neurophysiol 2012;108(4):1138-48.

16. Boonstra TA, Schouten AC, van der Kooij H. Identification of the contribution of the ankle and hip joints to multi-segmental balance control. J Neuroeng Rehabil 2013;10:23.

17. Shavelson RJ, Webb NM. Generalizability Theory. SAGE Publications, London; 1991.

18. Folstein MF, Folstein SE, McHugh PR. "Mini-mental state". A practical method for grading the cognitive state of patients for the clinician. J Psychiatr Res 1975;12(3):189-98.

19. Guralnik JM, Simonsick EM, Ferrucci L, Glynn RJ, Berkman LF, Blazer DG, et al. A short physical performance battery assessing lower extremity function: association with self-reported disability and prediction of mortality and nursing home admission. J Gerontol 1994;49(2):M85-M94.

20. Schouten AC, Boonstra TA, Nieuwenhuis F, Campfens SF, van der Kooij H. A bilateral ankle manipulator to investigate human balance control. IEEE Trans Neural Syst Rehabil Eng 2011;19(6):660-9.

21. Winter DA, Patla AE, Frank JS. Assessment of balance control in humans. Med Prog Technol 1990;16(1-2):31-51.

22. van der Kooij H, van AE, van der Helm FC. Comparison of different methods to identify and quantify balance control. J Neurosci Methods 2005;145(1-2):175-203.

23. de Vet HC, Terwee CB, Knol DL, Bouter LM. When to use agreement versus reliability measures. J Clin Epidemiol 2006;59(10):1033-9.

24. Fleiss RL. The design and analysis of clinical experiments. 1986.

25. Lariviere C, Ludvig D, Kearney R, Mecheri H, Caron JM, Preuss R. Identification of intrinsic and reflexive contributions to low-back stiffness: medium-term reliability and construct validity. J Biomech 2015;48(2):254-61.

26. Popovich JM, Jr., Reeves NP, Priess MC, Cholewicki J, Choi J, Radcliffe CJ. Quantitative measures of sagittal plane head-neck control: A test-retest reliability study. J Biomech 2015;48(3):549-54.

27. Reeves NP, Popovich JM, Jr., Priess MC, Cholewicki J, Choi J, Radcliffe CJ. Reliability of assessing trunk motor control using position and force tracking and stabilization tasks. J Biomech 2014;47(1):44-9.

28. Hendershot BD, Bazrgari B, Nussbaum MA, Madigan ML. Within- and between-day reliability of trunk mechanical behaviors estimated using position-controlled perturbations. J Biomech 2012;45(11):2019-22.
29. Wrisley DM, Stephens MJ, Mosley S, Wojnowski A, Duffy J, Burkard R. Learning effects of repetitive administrations of the sensory organization test in healthy young adults. Arch Phys Med Rehabil 2007;88(8):1049-54.

30. Moghadam M, Ashayeri H, Salavati M, Sarafzadeh J, Taghipoor KD, Saeedi A, et al. Reliability of center of pressure measures of postural stability in healthy older adults: effects of postural task difficulty and cognitive load. Gait Posture 2011;33(4):651-5.

31. Ageberg E, Roberts D, Holmstrom E, Friden T. Balance in single-limb stance in healthy subjects--reliability of testing procedure and the effect of short-duration sub-maximal cycling. BMC Musculoskelet Disord 2003;4:14.

32. Ford-Smith CD, Wyman JF, Elswick RK, Jr., Fernandez T, Newton RA. Test-retest reliability of the sensory organization test in noninstitutionalized older adults. Arch Phys Med Rehabil 1995;76(1):77-81.

33. Ruhe A, Fejer R, Walker B. The test-retest reliability of centre of pressure measures in bipedal static task conditions--a systematic review of the literature. Gait Posture 2010;32(4):436-45.

34. Santos BR, Delisle A, Lariviere C, Plamondon A, Imbeau D. Reliability of centre of pressure summary measures of postural steadiness in healthy young adults. Gait Posture 2008;27(3):408-15.

35. Asslander L, Peterka RJ. Sensory reweighting dynamics in human postural control. J Neurophysiol 2014;111(9):1852-64.

36. Weir JP. Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. J Strength Cond Res 2005;19(1):231-40.

37. Keating J, Matyas T. Unreliable inferences from reliable measurements. Aust J Physiother 1998;44(1):5-10.