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Reliability of assessing postural control during seated balancing using a physical human-robot interaction

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Abstract

This study evaluated the within- and between-visit reliability of a seated balance test for quantifying trunk motor control using input-output data. Thirty healthy subjects performed a seated balance test under three conditions: eyes open (EO), eyes closed (EC), and eyes closed with vibration to the lumbar muscles (VIB). Each subject performed three trials of each condition on three different visits. The seated balance test utilized a torque-controlled robotic seat, which together with a sitting subject resulted in a physical human-robot interaction (pHRI) (two degrees-of-freedom with upper and lower body rotations). Subjects balanced the pHRI by controlling trunk rotation in response to pseudorandom torque perturbations applied to the seat in the coronal plane. Performance error was expressed as the root mean square (RMSE) of deviations from the upright position in the time domain and as the mean bandpass signal energy (E_{mb}) in the frequency domain. Intra-class correlation coefficients (ICC) quantified the between-visit reliability of both RMSE and E_{mb} . The empirical transfer function estimates (ETFE) from the perturbation input to each of the two rotational outputs were calculated. Coefficients of multiple correlation (CMC) quantified the within- and between-visit reliability of the averaged ETFE. ICCs of RMSE and E_{mb} for all conditions were 0.84. The mean within- and between-visit CMCs were all 0.96 for the lower body rotation and 0.89 for the upper body rotation. Therefore, our seated balance test consisting of pHRI to assess coronal plane trunk motor control is reliable.

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Conflict of interest

Dr. N. Peter Reeves is an owner of a company (Sumaq Life LLC, East Lansing, MI) that develops technology, including robotic systems, to assess trunk control. None of the other authors of this manuscript have any conflicts of interest to disclose.

Keywords

Seated balance; Reliability assessment; Postural control; Lumbar spine

1. Introduction

Several sensory systems are involved in maintaining upright postural control and balance: visual, vestibular, and proprioception (Goodworth and Peterka, 2009; Maurer et al., 2006; Peterka, 2002). To determine the contribution of different sensory feedback pathways to trunk postural control, experiments are performed with sensory information either removed (e.g., closed eyes to remove visual information) (Goodworth and Peterka, 2009; Radebold et al., 2001) or corrupted (e.g., added vibration to reduce the quality of proprioceptive information) (Brumagne et al., 2004, 1999; Slota et al., 2008). Identifying the contributions of selected feedback pathways to trunk postural control is important for understanding the mechanisms of control and possible impairments. To that end, several studies have investigated the role of the trunk muscles in maintaining postural control using sitting balancing tests (Larivière et al., 2015; Oomen et al., 2015; Priess et al., 2015; Reeves et al., 2009; Sung et al., 2015; Van Daele et al., 2009; van Dieën et al., 2010a; Van Drunen et al., 2013; Willigenburg et al., 2013). Sitting posture has been selected, because the lower body is fixed so that ankle (Brumagne et al., 2008), knee, or hip (Mok et al., 2004) stabilization strategies are not possible; the movement is limited to the trunk to maintain balance.

For the seated balance test to be useful, it must reliably assess trunk control. Three studies addressed the reliability of seated balance tests, and the reported results range from poor to excellent (Cholewicki et al., 2000; Larivière et al., 2013; van Dieën et al., 2010b). These inconsistencies could stem from the fact that unperturbed postural sway was used as a measure of performance. In such a case, the postural sway (output signal) is transient behavior contaminated with neuromuscular noise, which is not repeatable. Thus, data reproducibility and test reliability may vary from one study to another. To overcome this problem, we developed a physical human-robot interaction (pHRI) system (Priess et al., 2015) that consists of a compliant unstable seat and, at the same time, provides torque perturbation as a known input to the system (Cruise et al., 2017; van Drunen et al., 2016, 2015, 2013). This pHRI also allows for the precise regulation of test difficulty by controlling rotational seat stiffness.

The purpose of this study was to assess the within- and between-visit reliability of a seated balance test under three conditions: eyes open (EO), eyes closed (EC), and eyes closed with lumbar muscle vibration (VIB). The reliability was evaluated using both time and frequency domain analysis, which capture the control attributes such as accuracy and speed. This study uses nonparametric analysis of input-output data, therefore, the contributions of feedback pathways (e.g., reflexive, and intrinsic) cannot be distinguished in this framework. Because the pHRI involved with this test was designed with a known input perturbation, we hypothesized that the performance measures in our seated balance test would be reliable.

2. Methods

2.1. Subjects

Thirty subjects participated in this study (Table 1). They were in self-reported good general health with no history of back pain lasting longer than three days or any neurological condition that could affect motor control. Subjects were instructed to wear their corrective lenses if their eyesight was impaired. Prior to testing, all subjects signed an informed consent. Michigan State University's Biomedical and Health Institutional Review Board approved the research protocol.

2.2. Data Collection

A pHRI was used in a seated balance test to assess trunk motor control in the coronal plane (Fig. 1). During the test, the robotic platform applied torque perturbations to the subjects. They were instructed to balance on the platform, with their arms crossed on the chest, using trunk motions in the coronal plane.

The pHRI consisted of a two degrees-of-freedom system (a one degree-of-freedom compliant-robot platform and a sitting subject) (Fig. 2 A). The first degree-of-freedom was the rotation of the subject's lower body (pelvis and below) together with the robot seat about the axis O and denoted by α_1 (lower-body angle). The second degree-of-freedom was the subject's upper body (trunk and above) rotation assumed to be about the fourth lumbar ($L4$) vertebra and denoted by α_2 (upper-body angle).

A torque τ generated by the robot's electric motor (C062C-13-3305, Kollmorgen, Radford, VA) was applied to the seat about O (Fig. 2 A). This torque contained components that are (1) a known perturbation $d(t)$, and (2) a stiffness-and-damping torque $\tau_{kb} = (k_s + b_s s)\alpha_1$ (Fig. 2 B). The perturbation signal $d(t)$ was a pseudorandom ternary sequence (Fig. 3 A).

The lower body angle (the seat angle about O) was measured by an encoder embedded in the motor. The angle was constrained to be within ± 12 degrees using mechanical stops. Our preliminary data demonstrated that the encoder accurately reflected the lower body angles. The RMS difference between the pelvis angles, captured independently with an optoelectronic motion system, and the encoder was 0.62 degrees. The upper body angle was measured by two string potentiometers (SP2-50, Celesco, Chatsworth, CA) attached to a chest harness and a belt at the fourth thoracic ($T4$) and the $L4$ levels, respectively (Fig. 1 B). The time series data of both angles is shown in Fig. 3 B–C for a single trial.

Each subject performed balancing tests under three conditions: EO, EC, and VIB. For each condition, subjects performed seven trials (four 30-second practice trials and three 30-second full trials). The last three (full) trials were used for analysis in this study. According to Larivière et al. (2013), three trials provided acceptable reliability.

To normalize test difficulty between subjects, both seat stiffness (k_s) and perturbation torque amplitude were adapted to the subject's ability before testing. The seat damping was empirically set to $b_s = 0.09 k_s - 2.5$ [Nms/rad] so that the seat was stable even when there was no subject.

First, the seat stiffness was normalized. The subject was instructed to balance on the seat with the stiffness provided from the electric motor, but without torque perturbations. Beginning from a high stiffness value ($k_s = 250$ Nm/rad), the subject was asked to maintain balance. If the subject successfully maintained balance (i.e., seat angle stayed within ± 12 degrees), ..., the operator gradually reduced the stiffness (similar to Tanaka et al., 2009) by 10 Nm/rad decrements until the subject could no longer maintain balance. This level of stiffness is called critical stiffness (Tanaka et al., 2009). The procedure was repeated once again and the lowest critical stiffness was chosen as a coarse estimate of the subject's critical stiffness.

For fine estimate of critical stiffness, the procedure was repeated three times starting from the coarse estimate of critical stiffness $+50$ Nm/rad. In these repetitions, the stiffness was reduced by 5 Nm/rad decrements until the subject could no longer maintain balance. The average of the three results was selected as a fine estimate of the critical stiffness. Thereafter, the experimental seat stiffness was set to twice the fine critical stiffness.

Second, the perturbation torque amplitude was normalized. With the experimental stiffness provided from the electric motor, a square-wave pulse torque with 1 Nm amplitude was applied to the seat for 1 s and then released. The pulse torque direction was random so that the subject could not predict it. If the seat angle did not reach ± 8 degrees after the pulse, the operator increased the amplitude by 1 Nm and the pulse torque was repeated until the seat angle reached approximately ± 8 degrees. This result was a coarse estimate of the perturbation torque amplitude.

Fine normalization of the torque amplitude was performed because the experimental trials used an input signal that was not a simple step input as the one used in the coarse normalization. Two normalization trials were performed for this purpose with an input signal that was very similar to the experimental trial signal with the maximal perturbation torque amplitude equal to the coarse estimate. The duration of a normalization trial was 30 s. The subject performed these trials with eyes closed and arms crossed. If the subject lost balance (seat angle ± 12 degrees identified with electronic switches) during a normalization trial, the perturbation torque amplitude was decreased by 1 Nm and the trial was repeated. If the subject passed two normalization trials, the highest passing perturbation torque amplitude was selected to be used in the experimental trials.

All data was logged using a data acquisition board (PCI-DAS6036, Measurement Computing, Norton, MA) and a quadrature encoder board (PCI-QUAD04, Measurement Computing, Norton, MA). The data was sampled at a rate of 1000 samples/s. Testing was conducted on three different visits, separated by a minimum of 23 hours. The test order was consistent between visits: (1) EO, (2) EC, and (3) VIB. Rest periods (approximately 30 s) were given between all trials.

2.3. Data Analysis

A simplified block diagram of the closed-loop system representing the seated balance test that utilizes pHRI is shown in Fig. 2 B. The plant P is a dynamical system that models the two degree-of-freedom seat/subject dynamics including the gravity effect (Fig. 2 A). Trunk

motor control generates lumbar torque about $L4$ and the robot generates total torque $\tau(t) = \dot{d}(t) - k_s a_1(t) - b_s \dot{a}_1(t)$ about O . The perturbation input is $d(t)$, while the system's output is $y(t) = [a_1 \ a_2]^T$.

The overall pHRI performance was characterized by two measures. First, the performance error, in the time domain, was expressed as a root mean square (RMSE) of deviations from the upright position (both angles equal zero). Since the time-domain data includes high-frequency noise, it is beneficial to consider the error in the frequency domain within a passband frequency region to which motor control mechanisms contribute. Therefore, the second performance measure was the mean bandpass signal energy of the error (E_{mb}) in the frequency domain.

The RMSE is a common performance measure in the time domain to assess motor control (Lee and Granata, 2008; Popovich et al., 2015; Reeves et al., 2014; van Dieën et al., 2010b). The error or performance error (e) used in the past studies refers to the scalar output assuming that subjects try to minimize it during a motor control test. For example, Popovich et al. (2015) and Reeves et al. (2014) used models with a single output, therefore, the scalar output was used as the performance error $e = y$. In our model, there are two outputs $y = [a_1 \ a_2]^T$. Thus, the motor control performance may be quantified by two errors, i.e. $e_1 = a_1$ and $e_2 = a_2$.

Frequency response was used to obtain additional performance measures of the test. From the frequency-domain data of the error e , we selected mean bandpass signal energy (E_{mb}) as the second performance measure. It was introduced in (Popovich et al., 2015) as

$$E_{mb, \alpha_i} = \frac{1}{k_h - k_l} \sum_{k=k_l}^{k_h} S_{\alpha_i \alpha_i}(f_k) \Delta f ; i=1, 2, \quad (1)$$

where $S_{a_i a_i}(f_k)$ is the power spectral density of the performance error $e_i = a_i$ at frequency f_k , Δf is the frequency resolution, and k_l and k_h are the indices of the lowest and highest frequency points in the passband region, respectively. The passband region was defined as the largest contiguous frequency range containing >3% of the maximum input power spectral density (Fig. 4 A) (Popovich et al., 2015). For these tests, the passband region was 0.1 – 1.6 Hz. The error power spectral density of a trial is depicted in Fig. 4 B–C. The

summation $\sum_{k=k_l}^{k_h} S_{\alpha_i \alpha_i}(f_k) \Delta f$ represents the area under the curve.

Intra-class correlation coefficients (ICC(C,k)) (McGraw and Wong, 1996) along with 95% confidence intervals (CI) were computed as a measure of the reliability between the three visits. ICC was computed for RMSE and E_{mb} using the mean of the last three trials of each visit. ICC was classified as fair (0.40–0.59), good (0.60–0.74), and excellent (0.75–1.00) (Cicchetti, 1994). The standard error of measurement (SEM) was calculated as $SD \sqrt{1 - ICC}$.

Coefficient of multiple correlations (CMC) was used to quantify within- and between-visit reliability. CMC was modified from (Kadaba, 1989) to use the frequency-domain data for

the passband region. For this purpose, the empirical transfer function estimates (ETFE) (Billinger, 2001; Ljung, 1999) were obtained from d to α_1 (Fig. 5 A–B and Fig. 6 A–B) and from d to α_2 (Fig. 5 C–D and Fig. 6 C–D). The three-dimensional ETFE (real, imaginary, and frequency dimensions) was used instead of the time-domain data used in (Kadaba, 1989).

Performance improvement across the three visits was assessed using a repeated-measures ANOVA. P-values ≤ 0.05 were considered significant. If the p-value for a certain condition (EO, EC, VIB) was significant, a post-hoc paired t-test (without corrections) was used for pairwise comparisons between visits.

3. Results

The ICCs of the performance measures presented in Table 2 and Table 3 are ≥ 0.84 , which reflect excellent between-visit reliability, however, ICCs associated with α_1 were better than that of α_2 . The confidence intervals have small ranges as well. The relatively smaller SEM in the E_{mb} case might be a result of the averaging in the ETFE computation. With respect to the overall characteristics of pHRI behavior in the frequency-domain, CMCs of within-visit reliability for ETFE were between 0.93 and 0.97, while CMCs of between-visit reliability ranged from 0.89 to 0.96 (Table 4). Both reflect excellent reliability, however, CMCs associated with α_1 were better than that of α_2 .

Comparing subject's performance over the three visits indicates that some learning in the selected conditions and measures may have occurred (Table 5 and Table 6). The time domain performance measure ($RMSE_{\alpha_1}$) improved across the three visits in the EO and VIB conditions ($p=0.05$ and 0.02 , respectively), whereas the frequency domain performance measure (E_{mb,α_1}) improved across the visits for only the VIB condition ($p=0.05$). Neither the time nor frequency domain performance measures for the EC condition changed significantly across the three visits ($p>0.05$). $RMSE_{\alpha_2}$ and E_{mb,α_2} did not have significant changes either.

The post-hoc pairwise comparisons revealed that $RMSE_{\alpha_1}$ was significantly smaller in visits 2 and 3 in comparison to visit 1 for the VIB condition. E_{mb,α_1} was significantly smaller in visit 3 in comparison to visit 1 for the VIB condition as well. For the EO condition, $RMSE_{\alpha_1}$ of visit 2 was significantly smaller than that of visit 1.

4. Discussion

The objective of this study was to determine if seated balance performance is reliable within and between testing sessions. Three conditions (EO, EC, and VIB) that could be used to examine various sensory pathways involved in postural control were assessed for reliability. The results of this study show that the seated balance test produced reliable performance measures under the three conditions, both in the time-domain (RMSE, ICC) as well as the frequency-domain (E_{mb} , CMC). When comparing with other studies, there is some evidence to suggest that the pHRI helps improve reliability. For instance, physical springs were used to adjust the rotational stiffness of a balance seat (Lee and Granata, 2008). Their reliability was poor to fair (ICC ranged 0.29–0.56). The difference in reliability between that study and

the present study may be explained by the presence of a perturbation signal. With the pHRI and the use of a perturbation signal, the response of trunk control, reflecting the system output, will be correlated to the input signal. This correlation improves the measurement reproducibility, which in turn improves reliability. For a seated balance test without a perturbation input, the output of the system is dominated by random noise generated within the neuromuscular system. Other studies that performed seated balance on a hemisphere without a perturbation signal had lower reliability (within-visit ICC range 0.03–0.70) (van Dieën et al., 2010b) than our pHRI with perturbation input, which supports the use of a pHRI along with perturbations when assessing trunk control.

Another advantage of the pHRI is the ability to easily regulate test difficulty, which also appears to be important for producing reliable performance measures of trunk control (Lee and Granata, 2008) and identifying differences between people with and without back pain (Radebold et al., 2001). Other seated balance set-ups have the ability to regulate test difficulty by changing the radius of the hemisphere (Cholewicki et al., 2000; Radebold et al., 2001; Reeves et al., 2008; van Dieën et al., 2010b) or by changing the position of the springs used to provide rotational stiffness (Larivière et al., 2015, 2013; Lee and Granata, 2008; Oomen et al., 2015). This adjustment, however, often requires the subject to be removed from the seat, and in the case of the hemisphere, has less resolution in changing the test difficulty. With the pHRI set-up, there is greater resolution in changing test difficulty and seat stiffness can be adjusted with the subject on the seat.

The results showed that some learning took place in the EO ($\text{PRMSE}_{\alpha_1} = 0.05$) and VIB conditions ($\text{PRMSE}_{\alpha_1} = 0.02$, $p_{E_{mb\alpha_1}} = 0.05$), which suggest that more practice may be required prior to assessing trunk control. Future work should investigate how much practice is needed to converge on relatively stable performance. Moreover, learning may be due to modulations in the contributing pathways (van Drunen et al., 2015) (e.g., intrinsic and vestibular pathways in the VIB condition). Future work should investigate such pathways by using parametric models.

One of the limitations of the study is that seat stiffness and torque perturbations were not normalized to the subject anthropometrics. The seat stiffness and the perturbation amplitude were determined based on the subject's performance during the normalization phase of data collection (i.e., critical seat stiffness and maximal perturbation amplitude at which the subject could maintain balance). Because the seated balance performance measures were reliable within and between visits, the current normalization protocol is suitable for repeated measures testing, such as assessing changes pre-post intervention. However, it is unclear if this normalization protocol could be used to assess between-group differences in trunk control, such as comparing trunk control between people with and without back pain. To allow for such comparisons, it may be necessary to normalize test difficulty to subject anthropometrics (i.e., dynamics from body size) and not overall balance performance (i.e., dynamics from body size and control capabilities). Future work should investigate various methods for normalizing test difficulty to anthropometrics, thus isolating the control element and allowing pHRI to be used more broadly to study trunk control in various populations.

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Fig 1.

The seated balance test utilizes a physical human-robot interaction (pHRI). (A) The upright equilibrium position. (B) A tilted equilibrium position.

Fig 2.

(A) The two degree-of-freedom physical model of the seated balance test utilizing a pHRI.

(B) Block diagram of the pHRI system. The input $d(t)$ is a torque signal perturbation, whereas the output $y(t) = [\alpha_1 \ \alpha_2]^T$ is the displacement vector (lower and upper body angles).

P - seat/subject plant dynamics, k_s -seat stiffness, b_s -seat damping, τ -total seat torque.

Fig 3.

The time series input-output data of a single trial. (A) The perturbation torque $d(t)$ is a pseudorandom ternary sequence of 5 Nm amplitude. (B, C) The lower and upper body angles (α_1 , α_2), respectively.

Fig 4.
The power spectral density S_{xx} of (A) the perturbation d , (B) the error $e_1 = a_1$ and (C) the error $e_2 = a_2$ of a trial, respectively.

Fig 5.

The frequency domain data represented in amplitude and phase of the ETFE for the same trial/subject/condition across three visits. The curves are plotted in the passband region of 0.1 – 1.6 Hz. (A, B) The ETFE from d to α_1 . (C, D) The ETFE from d to α_2 .

Fig 6.

The frequency domain data represented in amplitude and phase of the ETFE for the same trial/condition/visit across three subjects. The curves are plotted in the passband region of 0.1 – 1.6 Hz. (A, B) The ETFE from d to α_1 . (C, D) The ETFE from d to α_2 .

Table 1

Characteristics of the subjects presented as mean $\pm$ standard deviation.

	Females (n=19)	Males (n=11)
Height (m)	1.63 $\pm$ 0.05	1.78 $\pm$ 0.07
Weight (kg)	70.6 $\pm$ 12.8	86.1 $\pm$ 11.5
Age (years)	34.7 $\pm$ 12.0	30.7 $\pm$ 12.3

Table 2

Between-visit reliability (ICC(C, k)) of seated balance test performance measures in time domain (RMSE). ICC is given as mean (95% CI, SEM).

Condition	ICC of RMSE _{a1}	ICC of RMSE _{a2}
EO	0.98(CI 0.96 – 0.99, SEM 0.11deg)	0.89(CI 0.80 – 0.94, SEM 0.14deg)
EC	0.98(CI 0.96 – 0.99, SEM 0.10deg)	0.88(CI 0.78 – 0.94, SEM 0.13deg)
VIB	0.98(CI 0.96 – 0.99, SEM 0.10deg)	0.88(CI 0.79 – 0.94, SEM 0.14deg)

Table 3

Between-visit reliability (ICC(C, k)) of seated balance test performance measures in frequency domain (E_{mb}). ICC is given as mean (95% CI, SEM).

Condition	ICC of E_{mb,a_1}	ICC of E_{mb,a_2}
EO	0.98(CI 0.96 – 0.99, SEM 0.012deg ²)	0.86(CI 0.75 – 0.93, SEM 0.007deg ²)
EC	0.98(CI 0.96 – 0.99, SEM 0.012deg ²)	0.84(CI 0.71 – 0.92, SEM 0.007deg ²)
VIB	0.98(CI 0.96 – 0.99, SEM 0.011deg ²)	0.86(CI 0.74 – 0.93, SEM 0.008deg ²)

Table 4

Within- and between-visit reliability expressed as CMC of the empirical transfer function estimates (ETFE) from perturbation torque d to lower and upper body angles (a_1 and a_2 , respectively) in frequency domain. CMC values are given as mean $\pm$ standard deviation.

Condition	Within - visit CMC		Between – visit CMC	
	$d \rightarrow a_1$	$d \rightarrow a_2$	$d \rightarrow a_1$	$d \rightarrow a_2$
EO	0.97 ± 0.02	0.94 ± 0.05	0.96 ± 0.02	0.89 ± 0.11
EC	0.97 ± 0.01	0.93 ± 0.05	0.96 ± 0.02	0.89 ± 0.07
VIB	0.97 ± 0.02	0.93 ± 0.05	0.96 ± 0.02	0.91 ± 0.06

Table 5

RMSE across visits and conditions. Values are given as mean ± standard deviation.

Condition	RMSE _{α₁} (deg)			P-value	RMSE _{α₂} (deg)			P-value
	Visit 1	Visit 2	Visit 3		Visit 1	Visit 2	Visit 3	
EO	3.34 ±0.79	3.22 [#] ±0.77	3.29 ±0.77	0.05	1.29 ±0.51	1.20 ±0.46	1.27 ±0.47	0.29
EC	3.41 ±0.70	3.33 ±0.74	3.33 ±0.71	0.14	1.26 ±0.45	1.20 ±0.39	1.23 ±0.44	0.66
VIB	3.40 ±0.69	3.30 [#] ±0.72	3.28 [#] ±0.69	0.02	1.33 ±0.48	1.28 ±0.46	1.32 ±0.46	0.71

[#] indicates values significantly smaller in comparison to Visit 1 (p 0.05).

Table 6

E_{mb} across visits and conditions. Values are given as mean $\pm$ standard deviation.

Condition	$E_{mb,\alpha_1}(\text{deg}^2)$			$E_{mb,\alpha_2}(\text{deg}^2)$			P-value
	Visit 1	Visit 2	Visit 3	Visit 1	Visit 2	Visit 3	
EO	0.197 $\pm$ 0.082	0.184 $\pm$ 0.077	0.192 $\pm$ 0.081	0.033 $\pm$ 0.023	0.028 $\pm$ 0.023	0.031 $\pm$ 0.022	0.37
EC	0.199 $\pm$ 0.077	0.194 $\pm$ 0.078	0.194 $\pm$ 0.074	0.030 $\pm$ 0.020	0.027 $\pm$ 0.017	0.029 $\pm$ 0.021	0.57
VIB	0.198 $\pm$ 0.076	0.189 $\pm$ 0.078	0.187 [#] $\pm$ 0.069	0.034 $\pm$ 0.024	0.031 $\pm$ 0.024	0.033 $\pm$ 0.024	0.70

[#] indicates values significantly smaller in comparison to Visit 1 (p 0.05).