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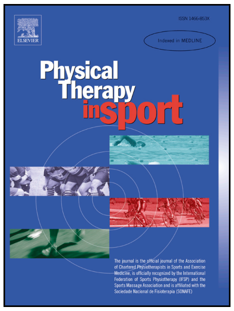

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Title

The within-day and between-day reliability of using sacral accelerations to quantify balance performance

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The within-day and between-day reliability of using sacral accelerations to quantify balance performance

Objectives: To investigate the between-day and within-day reliability of a sacral mounted accelerometer to quantify balance performance and different balance metrics.

Design: Experimental, cross-sectional.

Setting: Laboratorial experiment.

Participants: Thirty healthy volunteers.

Main outcome measures: Balance tasks were double leg stance, tandem stance and single leg stance with eyes open and closed. Performance was measured by converting accelerations into path length (PL, length of the sway trace), jerk (jerkiness of sway trace) and root mean square (RMS) of the accelerations.

Results: Within-day ICC for PL were excellent (mean 0.78 95%CI 0.68-0.89), with Jerk and RMS demonstrating means of 0.60 and 0.47 respectively. The mean percentage minimal detectable change (MDC) within-day were small for PL (mean 6.7%, 95%CI 5.3-8.1).

Between-day ICC were good for PL (mean 0.61, 95%CI 0.50-0.71), but more varied for Jerk and RMS. The mean percentage MDC was small for PL (mean 6.1%, 95%CI 5.0-7.2). No significant differences were determined for measurements between-days for any metric or task. PL had the highest discriminatory value between the 8 tasks.

Conclusions: The sacral mounted accelerometer reliably measured balance performance within- and between-days. The PL is the recommended metric as it was the most reliable, most discriminatory and most sensitive to change.

1. Introduction

Balance measurement has traditionally focused on the determination of postural sway quantified by tracking the trajectory of the centre of pressure (COP). This commonly requires the use of expensive laboratory mounted force plates (Mancini et al., 2012). Clinicians often measure balance using crude measures such as time spent on one leg or star excursion balance test (O'Sullivan et al., 2009; Coughlan et al., 2013). These measures provide limited detail about the quality of performance. Clinicians are therefore faced with a challenge of obtaining detailed objective information regarding balance performance without being constrained to a laboratory environment.

Body mounted sensors, such as accelerometers, have been suggested as an alternative balance measurement method (Moe-Nilssen 1998a; Moe Nilssen 1998b; Moe-Nilssen & Helbostad 2002). These sensors are capable of measuring linear acceleration along each sensing axis and when attached close to the body's centre of mass (COM) have the ability to measure acceleration of the body's COM. This has been suggested as a viable method to quantify balance (Moe-Nilssen 1998). Studies comparing the traditional force plate measures with the accelerometer method have shown promising results. However the two methods measure balance in unique ways. Force plates are usually employed to measure the behaviour of the COP, which represents the point location of the vertical ground reaction force vector (Winter 1995). However changes in COP do not always correspond to change in the position of the body's COM (Winter 1995). Accelerometers, on the other hand, measure acceleration of the COM and therefore describe the body's attempt to control movement of the COM (Adlerton & Moe-Nilssen 2003). As the two devices measure different metrics they are not necessarily interchangeable, however the correlation of the attempts to maintain balance are good (Mayagoitia et al., 2002; Whitney et al., 2011; Mancini et al., 2012). Therefore it may be possible to obtain sway signatures using the accelerometer method which has distinct advantages over the force plate method for the clinician being smaller, cheaper and not constrained to a specific environment.

There are a few technical issues around the use of accelerometers for measuring postural sway. Body-worn sensors are unlikely to be mounted perfectly to the horizontal and vertical resulting in an element of sensor tilt. This tilt affects the output of the accelerometer as acceleration due to gravity is an intrinsic component of the sensor output. This aspect needs addressing in order to resolve the true accelerations. This can be overcome by registering the degree of tilt of the sensor and using this to remove the gravity component of the sensor output. This method has been successfully applied in previous research involving accelerometers (Moe-Nilssen 1998a; Morgado-Ramirez et al., 2013; Williams & Cuesta-Vargas 2014) and accelerometers have successfully been used to quantify balance in older persons who fall (Doheny et al., 2012), children with dyslexia (Moe-Nilssen et al., 2003) and those with Huntington's disease (Dalton et al., 2013), Parkinson's disease (Mancini et al., 2011) and Vestibular disorders (Marchetti et al., 2013).

More recently these devices have been designed so that data is presented in a usable format for clinicians and as such could be used in every day practice. However, there is still a requirement to explore the reliability of these devices. Test-retest reliability for various stance tasks, including double leg stance (ICC 95% confidence interval (CI) = 0.35-0.89; Mancini et al., 2012), single leg stance (ICC 95%CI = 0.62-1; Moe-Nilssen 1998b) and tandem stance tasks (ICC 95% CI = 0.75-0.89) have been reported. These values represent the combination of biological (human) and equipment (sensor) variability and the spread of the confidence intervals demonstrate the inherently variability in human movement. Furthermore different specific metrics have been used to quantify performance from the acceleration data. Whitney et al (2011) demonstrated that the path length (PL), a measure of the length of the mediolateral acceleration data plotted against the anteroposterior acceleration data, was the most reliable measure across a range of balance tasks (ICC range 0.63-0.80). Mancini et al., (2012) also reported that PL was the most reliable balance performance metric however also suggested Jerk, the time derivative of acceleration, was reliable, a finding supported by Marchetti et al., (2013). In addition to PL and jerk previous authors have also reported the root mean square value (RMS) as a method of quantifying postural sway. Reliability estimates of 0.51-0.81 have been reported (Moe-Nilssen 1998b; Mancini et al., 2012) for double leg stance and single leg stance. It would appear that PL, Jerk and RMS are the most commonly reported metrics to quantify balance measured using an accelerometer, with previous studies demonstrating their reliability. Previous studies have not investigated these metrics across a wide range of balance tasks or investigated the between-day reliability of such a method. It is suggested that before such a method can be accepted in clinical practice a better understanding of the variability of repeated testing, within and between-day, is required along with the computation of the minimal detectable change (MDC). These values will then enable clinicians to interpret changes in balance performance to go beyond that expected from normal human variability.

The aim of this study was to determine the within-day and between-day repeated measures reliability of a novel device for measuring balance within a clinical setting, along with the determination of minimal detectable change values across a series of balance tasks.

2. **Methods**

This study employed an experimental cross-sectional test re-test study design.

2.1 Participants

Thirty participants were recruited from the student population within Bournemouth University. All participants were free from any musculoskeletal or neurological disorders or any other conditions which may affect their balance. Bournemouth University ethics committee granted the study ethical approval and all participants provided informed written consent to participate in the study. Mean (sd) age was 28.8 (8.7) years, height 1.71 (0.1) m, weight 73.4 (15.3) kg and 18 were female. A smaller sample of seven was asked to return the following day to repeat the tasks and explore the between-day reliability (mean (sd) age 24.9 (4.8) years, height 1.75 (11.5) m, weight 75.0 (15.3) kg). This value was calculated by declaring an acceptable correlation coefficient of 0.75, with alpha as 0.05 resulting in the required sample size, to achieve a power of 0.8, of 7 as calculated by Gpower (3.0).

2.2 Instrumentation

A commercially available balance sensor (THETAmetrix, Waterlooville, Hampshire, UK) was used to quantify balance. The sensor's dimensions were 73mm x 45mm x 19mm and weight 58 grams. The sensor houses a triaxial accelerometer and triaxial rate gyroscope which communicate wireless to a PC. The accelerometer measures linear acceleration along its sensing axes while the rate gyroscope quantifies rate of turn about its sensing axes. The company supplied software uses both sensing elements to overcome the limitations of using an accelerometer in isolation (outlined in the introduction), namely the dynamic correction for sensor tilt and removal of the gravity component of the signal. Therefore the software calculates orientation independent linear acceleration at 16Hz. With the sensor attached to the skin over the sacrum this acceleration data represents the small adjustments used to maintain balance i.e. the postural sway of the sacrum.

2.3 Procedure

Participants' height (Seca 274 Stadiometer, Seca, UK) and weight (Seca 761 Mechanical Scales, Seca, UK) were measured and the balance sensor was attached to the skin over the spinous process of S2 using double sided tape. This location was chosen as it is close to the centre of mass of the human body (Mancini et al., 2011; Whitney et al., 2011; Kim et al., 2013). All participants wore self-selected training shoes throughout as this reflects function and clinical practice. Eight balance tasks were completed, namely double leg stance, feet naturally apart with eyes open (DLSFNEO); double leg stance feet naturally apart with eyes closed (DLSFNEC); double leg stance, feet together with eyes open (DLSFTEO); double leg stance, feet together with eyes closed (DLSFTEC); tandem stance (right foot in front of the left) eyes open (TandEO); tandem stance eyes closed (TandEC); single leg stance with eyes open (SLSEO) and single leg stance with eyes closed (SLSEC). Participants were instructed to stand and maintain balance as best they can. Participants stood on a line positioned 2 m from the wall and asked to look at a wall marker 1.7m high during the eyes open tasks. The orders of the tests were randomised using opaque envelopes to minimise the potential effect of fatigue on performance. Each task was completed for 30 seconds and repeated 3 times with as much rest time as required by the participants between tests (typically 45s – 90s). If the participant shifted their feet during the task the data were discarded and trial repeated.

2.4 Data Analysis

Data were captured using the company provided balance sensor software. This converts the anterio-posterior (AP) and medio-lateral (ML) linear accelerations into three specific metrics quantifying sway behaviour of the pelvis. The three metrics were path length (PL), jerk and root mean square (RMS). Path length computes the length of the sway path created by the AP and ML accelerations. The length of the path between each sequential data point for AP acceleration (sample (x+1) – sample x) is calculated and summed. This is repeated for ML acceleration and the total path length is the sum of the AP path length and the ML path length. It is measured in mg (where m stands for milli- and g for units of gravity). Jerk represents the jerkiness or 'twitchiness' of the movement of the pelvis in mg^2/s (where m stands for milli-, g for units of gravity and s for seconds). It is computed by calculating the integral of the square of the gradient of acceleration and then normalised with respect to time. The total jerk is the sum of the jerk for AP accelerations and ML accelerations. RMS is the standard deviation of the acceleration-time series in mg (where m stands for milli- and g for units of gravity). These metrics were used to provide quantification of the balance performance for each task. An example of the output is presented in figure 1. All data were tested for normality using the

Shapiro-Wilk test and as the majority of data were not normally distributed non-parametric statistics were used throughout. Within-day repeated measures reliability was determined for the three trials using intra-class correlation coefficients (ICC 3, 1) and standard error of measurement (SEM) was calculated for each task (Rankin & Stokes 1998; Bruton et al., 2000). To aid clinical interpretation the minimal detectable change (MDC) was also computed and converted into a percentage of the median score to enable comparison. Significant differences between tasks were explored using Friedman's test with post hoc Wilcoxon (bonferoni corrected) where appropriate. *** place figure 1 about here ***

Three trials of each task were completed on the follow-up day (>24 hours post initial measurements) from which between day reliability was determined using the mean of the three measurements across the two days to compute an ICC (3, k) (Rankin & Stokes 1998). Additional SEM and percentage MDC for between day measurements were also calculated. Significant differences were explored between day 1 and day 2 measurements using Friedman's with post hoc Wilcoxon (Bonferoni corrected) and effect size calculations where appropriate.

3. Results

The median values and SEM for each balance metric during each task is presented in table 1.

3.1 Within day reliability

The ICC values ranges from poor to excellent across the different tasks and metrics. The PL consistently demonstrated the greatest ICC across all tasks with mean 0.78 and 95%CI 0.68 - 0.89 (table 1). RMS demonstrated moderate to good ICC values with mean 0.60 and 95%CI 0.47-0.72. Jerk consistently demonstrated the lowest ICCs across all tasks with a mean of 0.47 and 95% confidence interval of 0.28 – 0.66 (table 1).

The percentage MDCs were low for PL (mean 6.7; 95%CI 5.3-8.1) and significantly higher for JERK and RMS. These results suggest that PL represents a measure that is more sensitive to detect change in performance than Jerk or RMS.

There were significant differences between the balance tasks for PL ($\chi^2 = 513.3$, $p < 0.01$). Post hoc analysis using Wilcoxon sign rank tests (with Bonferoni correction) showed significant differences between TandEO and all others; TandEC and all others; SLSEO and all others; SLSEC and all others, as well as DLSFNEO and DLSFTEC; DLSFNEC and DLSFTEC for PL (figure 2a). There were significant differences between tasks for Jerk ($\chi^2 = 396.1$, $p < 0.01$). Post hoc analysis showed significant differences between TandEO, TandEC, SLSEO and SLSEC and all other tasks (figure 2b). There were significant differences between tasks for RMS ($\chi^2 = 177.4$, $p < 0.01$). Post hoc analysis showed significant differences between TandEO and all other tasks except DLSFTEO and SLSEO; TandEC and all other tasks; SLSEO and all other task except DLSFTEO and TandEO; SLSEC and all other tasks (figure 2c). These results indicate that the double leg stance tasks differ little in the actual resultant metrics suggesting they challenge balance to a similar amount. Tandem stance tasks and single leg stance tasks yield significantly different results therefore provides a unique degree of balance challenge. These results also demonstrate that PL is more sensitive in discriminating between different balance tasks. *** place table 1 about here *** *** place figure 2 about here ***

3.2 Between-day reliability

The ICC values ranged from poor to excellent across the different tasks and metrics (table 2). The PL consistently demonstrated the greatest ICC across tasks with mean 0.61 and 95% CI 0.50 - 0.71. Jerk and RMS displayed much greater variability across tasks with mean ICC 0.33 with 95% CI 0.09 – 0.58 and 0.35 with 95% CI 0.17 – 0.52 respectively (table 2). Greatest between day reliability was determined for the SLSEC task across all metrics demonstrating that despite this task being the most challenging it was also the most consistent (table 2).

The percentage MDC values were small especially for PL (mean 6.1; 95%CI 5.0 – 7.2). The SLSEO presented the smallest percentage MDC value, demonstrating that this task may be the most sensitive to detect change across days.

Using a Bonferoni corrected alpha value of 0.007 there were no significant differences between day 1 and day 2 measurements for any of the tasks or variables (PL $p = 0.11 – 1.0$; jerk $p = 0.05 – 1.0$; RMS $p = 0.03 – 1.0$).

4. Discussion

The aim of this study was to determine the reliability of a balance sensor based on accelerometry as well as obtain MDCs and explore differences between balance tasks. Previous studies have suggested the use of accelerometers to measure balance, however previous studies have not explored the range of tasks, between-day reliability or MDCs as in this study.

The within-day repeated measures reliability results from this study suggest that the balance sensor can be used with confidence as ICC values were generally good to excellent. Previous studies have used triaxial accelerometers (Moe-Nilssen 1998b), dual-axial accelerometers (Whitney et al., 2011) or an inertial sensor (Mancini et al., 2012) in their measurement of balance. Published studies using RMS have demonstrated ICCs of 0.16 – 0.71 for DLSFNEO (Moe-Nilssen 1998b; Whitney et al., 2011, Mancini et al., 2012), 0.46 - 0.52 for DLSFNEC (Moe-Nilssen 1998b; Whitney et al., 2011) and 0.81 for SLSEO (Moe-Nilssen 1998b). As with the current study greater reliability has been reported for PL ranging from 0.72 – 0.89 for DLSFNEO and 0.72 for DLSFNEC (Whitney et al., 2011; Mancini et al., 2012). Therefore the results of the current study are in agreement with those previously reported and the findings for the functional tasks not previously studied are also similar. However, like those previous studies not all metrics in the current study performed the same. Total PL demonstrated the greatest reliability and smallest MDC, outperforming the other metrics significantly. The MDC is an important metric for clinicians to use (Donoghue et al., 2009). This metric provides an outcome measure for an intervention, or may be used to track deterioration in a condition, such as recurrent ankle sprain or Diabetes. These results therefore suggest a change in total PL of 10% could be considered true clinical change as opposed to variability in task performance for any of the tasks measured in this study. Furthermore a change >5.6% for single leg stance task could be considered true change. Other performance metrics (Jerk and RMS) showed more variability in the MDC values. It is therefore suggested that when using these metrics, attention should be paid to the individual scores for individual tasks, a finding supported in the literature (Whitney et al., 2011).

This is this first time data regarding performance of these tasks has been reported. Performance scores for a particular balance task could be compared the values in table 2 to determine level of balance function or impairment. This information can then in-turn guide clinical decision making. Caution is advised as the sample in this study was limited to young healthy adults making extrapolation to other populations questionable. This study determined that some tasks yielded significantly different results to others aiding our understanding of the underlying constructs of the tasks. There were no differences in the DLSFN and DLSFT tasks for jerk and RMS, however there were differences for DLSFTEC using total PL. This suggests total PL had the greatest discriminatory ability, however the results may differ in those with balance impairments. The tasks of TandEO, TandEC, SLSEO and SLSEC resulted in a different score from each other and all other tasks, suggesting that these tasks explore a unique level of balance challenge and could be considered important in a battery of balance testing. Conversely as little differences between the double leg stance tasks were determined it is possible these tasks provide the same level of balance challenge, which in an unimpaired population, is minimal suggesting that the inclusion of all these tasks may be unnecessary. However the level of balance challenge and subsequent results may differ in those with balance impairments.

The between day reliability results show poorer ICC scores suggesting less consistency across the two days. The PL again demonstrates greater between day reliability with all ICC values greater than 0.4, however the majority were around 0.6 or more. Some between day reliability scores were very low for jerk and RMS suggesting an unreliable metric and caution should be advised during their use. This is the first time that between day reliability of accelerometery for balance measurement has been reported and represents a novel finding in the literature. In order to determine if a measurement method has sufficient reliability, exploration beyond the ICC is required. Indeed if there is a lack of range in the values reported or put another way, results from all participants are clustered together, the ICC will be artificially low (Moe-Nilssen 1998b), therefore it may be more clinically meaningful to use the MDC values to determine if the measurement is ‘reliable enough’ for a specific application. The between day percentage MDC values were very small for PL suggesting that clinicians could be confident with changes in performance of >7% represent true change across days. The same was not found for jerk and RMS where much greater changes would be necessary to denote true change.

The current study outlines the use of a quick and simple method for objectively measuring balance. Attachment of the device takes less than a minute and the software produces an immediate report detailing the individual’s performance making the quantification of balance quick and easy. A sacral mounted balance sensor enables immediate assessment of balance in any environment, removing the reliance on expensive laboratory systems. Other platform based systems have limitations regarding footwear, such as studs or cleats, whereas a sacral mounted sensor could quickly quantify the athletes balance pitch-side, for example in response to head injury. It is believed that as therapy and rehabilitation continue to evolve to demand greater objectivity and more specific outcome measures, such instrumentation is well placed to fulfil that need. It is recommended that the PL total be used by clinicians as a single score from which to quantify balance as this has shown to offer the greatest within-day and between-day reliability, best discriminatory ability and lowest MDC making it the most sensitive to change.

This study was limited to young healthy individuals therefore the results in different populations are likely to be different. All tasks were designed to test static balance and as such results will differ for more dynamic balance tasks. Results will also differ for functional tasks which were not explored in the current study. Participants in this study used their own shoes therefore results may not reflect performance barefoot. It is also important to note that sacral accelerometry cannot take into consideration the influence of muscle and the effects muscle contraction can have on the centre of pressure. Therefore it is recognised that the measurement of centre of pressure and measurement of acceleration of the pelvis are not the same.

5. Conclusion

This study demonstrates that a sacral mounted balance sensor can reliably measure postural sway across a range of balance tasks, both within day and between days. Tandem stance and single leg stance tasks produce significantly different scores to the other tasks and should be included in future balance testing. The PL is the most reliable metric of balance function and has the lowest minimal detectable change values making it the most sensitive to change. It is recommended that PL be used as the key variable for clinical assessment of balance using a balance sensor as described in this study.

6. References

Adlerton, A.N., Moritz, U., Moe-Nilssen, R. (2003). Forceplate and accelerometer measures for evaluating the effect of muscle fatigue on postural control during one-legged stance. Physiotherapy Research International. 8(4): 187-199.

Bruton, A., Conway, J.H., Holgate, S.T. (2000). Reliability: what is it, and how is it measured? Physiotherapy. 86(2): 94-99.

Dalton, A., Khalil, H.,Busse, M., Rosser, A., van Deursen, R., Olaighin, G. (2013). Analysis of gait and balance through a single triaxial accelerometer in presymptomatic and symptomatic Huntington's disease. Gait & Posture. 37(1): 49-54.

Doheny, E.P., McGrath, D., Greene, B.R., Walsh, L., McKeown, D., Cunningham, C., Crosby, L., Kenny, R.A., Caulfield, B. (2012). Displacement of centre of mass during quiet standing assessed using accelerometry in older fallers and non-fallers. Conference Proceedings IEEE Engineering in Medicine and Biology Society. 2012:3300-3303.

Donoghue, D., Physiotherapy Research and Older People group., Stokes E.K. (2009). How much change is true change? The minimal detectable change of the berg balance scale in elderly people. Journal of Rehabilitation Medicine. 41(5): 343-346.

Kim, S-C., Kim, M.J., Kim, N., Hwang, J.H., Han, G.C. (2013). Ambulatory balance monitoring using a wireless attachable three-axis accelerometer. Journal of Vestibular Disorders. 23: 217-225.

Mancini, M., Salarian, A., Carlson-Kuhta, P., Zampieri, C., King, L., Chiari, L., Horak, F.B. (2012). ISway: a sensitive, valid and reliable measure of postural control. Journal of NeuroEngineering and Rehabilitation, 9: 59.

Mancini, M., Horak, F.B., Zampieri, C., Carlson-Kuhta, P., Nutt, J.G., Chiari, L. (2011). Trunk accelerometry reveals postural instability in untreated Parkinson's disease. Parkinsonism and Related Disorders. 17: 557-562.

Marchetti, G.F., Bellanca, J., Whitney, S.L., Lin, J.C-C., Musolino, M.C., Furman, G.R., Redfern, M.S. (2013). The development of an accelerometer-based measure of human upright static anterior-posterior postural sway under various sensory conditions: Test-retest reliability, scoring and preliminary validity of the Balance Accelerometry Measure (BAM). Journal of Vestibular Research. 23: 227-235.

Mayagoitia, R.E., Lotters, J.C., Veltink, P.H., Hermens, H. (2002). Standing balance evaluation using a triaxial accelerometer. Gait & Posture. 16:55-59.

Moe-Nilssen, R., (1998a). A new method for evaluating motor control in gait under real-life environmental conditions. Part 1: The instrument. Clinical Biomechanics. 13(4-5): 320-327.

Moe-Nilssen, R. (1998b). Test-retest reliability of trunk accelerometry during standing and walking. Archives of Physical Medicine and Rehabilitation. 79: 1377-1385.

Moe-Nilssen, R., Helbostad, J.L. (2002). Trunk accelerometry as a measure of balance control during quiet standing. Gait & Posture. 16(1): 60-68.

Moe-Nilssen, R., Helbostad, J.L., Talcott, J.B., Toennessen, F.E. (2003). Balance and gait in children with dyslexia. Experimental Brain Research. 150:237-244.

Morgado-Ramirez, D.Z., Strike, S., Lee, R.Y. (2013). Measurement of transmission of vibration through the human spine using skin mounted inertial sensors. Medical Engineering in Physics. 35(5):690-695.

O'Sullivan, M., Blake, C., Cunningham, C., Boyle, G., Finucane, C. (2009). Correlation of accelerometry with clinical balance tests in older fallers and non-fallers. Age and Aging. 38: 308-313.

Rankin, G., Stokes, M. (1998). Reliability of assessment tools in rehabilitation: an illustration of appropriate statistical analyses. Clinical Rehabilitation. 12: 187-199.

Whitney, S.L., Roche, J.L., Marchetti, G.F., Lin, C-C., Steed, D.P., Furman, G.R., Musolino, M.R., Redfern, M.S. (2011). A comparison of accelerometry and centre of pressure measures during computerized dynamic posturography: A measure of balance. Gait & Posture. 33: 594-599.

Winter, D.A. (1995). Human balance and posture control during standing and walking. Gait & Posture. 3(4): 193-214.

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Figure Captions

Figure 1. Example output from balance sensor.

Figure 2. Median Path Length (mg) (a); Jerk (mg^2/s)(b); Root Mean Square (mg) (c) values for each task with results of significance testing. DLSFNEO, double leg stance feet natural eyes open; DLSFNEC, double leg stance feet natural eyes closed; DLSFTEO, double leg stance feet together eyes open; DLSFTEC, double leg stance feet together eyes closed; TandEO, Tandem stance eyes open; TandEC, Tandem stance eyes closed; SLSEO, Single leg stance eyes open; SLSEC, Single leg stance eyes closed; PL, path length.

Table 1. Median scores for PL, jerk and RMS for each balance task and within day reliability and percentage minimal detectable change.

Task		PL (mg)	Jerk (mg^2/s)	RMS (mg)
DLSFNEO	**Median**	141.1	5.8	3.3
	ICC	0.85	0.61	0.74
	SEM	8.3	7.2	0.7
	%MDC	5.7	128.3	72.5
DLSFNEC	**Median**	142.1	6.1	2.9
	ICC	0.88	0.48	0.77
	SEM	6.8	13.5	1.0
	%MDC	5.1	167.0	96.1
DLSFTEO	**Median**	150.7	7.6	4.1
	ICC	0.44	0.09	0.42
	SEM	31.9	18.6	1.7
	%MDC	10.4	157.2	88.2
DLSFTEC	**Median**	156.0	8.3	3.5
	ICC	0.90	0.59	0.77
	SEM	8.1	6.0	0.9
	%MDC	5.1	81.9	75.9
TandEO	**Median**	189.3	11.5	4.2
	ICC	0.69	0.03	0.26
	SEM	27.3	74.9	3.3
	%MDC	7.7	208.6	120.7
TandEC	**Median**	338.6	26.2	5.6
	ICC	0.85	0.73	0.52
	SEM	111.5	66.9	1.6
	%MDC	8.6	86.5	62.3
SLSEO	**Median**	247.5	17.3	4.8
	ICC	0.84	0.52	0.67
	SEM	23.8	15.6	1.3
	%MDC	5.5	63.3	66.1
SLSEC	**Median**	766.5	133.8	6.6
	ICC	0.80	0.73	0.63
	SEM	244.1	229.2	2.5
	%MDC	5.6	31.4	66.3

DLSFNEO, double leg stance feet natural eyes open; DLSFNEC, double leg stance feet natural eyes closed; DLSFTEO, double leg stance feet together eyes open; DLSFTEC, double leg stance feet together eyes closed; TandEO, Tandem stance eyes open; TandEC, Tandem stance eyes closed; SLSEO, Single leg stance eyes open; SLSEC, Single leg stance eyes closed; PL, path length; RMS, root mean squared; ICC, intraclass correlation coefficient; SEM, standard error of measurement; MDC, minimal detectable change.

Table 2. Absolute mean difference (sd) between participants day one and day two for each of the tasks with between day ICC, SEM and percentage MDC values.

Task		PL (mg)	Jerk (mg^2/s)	RMS (mg)
DLSFNEO	**AbsMeanDiff**	26.2	15.9	1.2
	sd	20.3	30.6	1.3
	ICC	0.44	0.27	0.36
	SEM	15.2	26.1	1.0
	%MDC	7.0	206.4	86.7
	Effect size	0.52	0.41	0.53
DLSFNEC	**AbsMeanDiff**	18.2	7.0	1.3
	Sd	17.9	7.4	1.1
	ICC	0.57	0.21	0.15
	SEM	11.8	6.6	1.0
	%MDC	6.7	86.6	72.8
	Effect size	0.54	0.80	0.89
DLSFTEO	**AbsMeanDiff**	29.3	6.1	0.9
	Sd	32.0	6.9	0.6
	ICC	0.57	0.43	0.61
	SEM	20.9	5.2	0.4
	%MDC	8.2	66.8	37.9
	Effect size	0.09	-0.07	0.09
DLSFTEC	**AbsMeanDiff**	21.9	28.9	1.5
	Sd	16.6	38.8	0.9
	ICC	0.57	0.07	0.45
	SEM	10.8	37.4	0.7
	%MDC	5.3	139.0	48.8
	Effect size	0.09	0.34	0.10
TandEO	**AbsMeanDiff**	21.4	17.9	2.0
	Sd	25.0	18.0	1.5
	ICC	0.70	0.04	0.29
	SEM	13.6	17.6	1.3
	%MDC	6.1	122.5	32.5
	Effect size	-0.25	-0.28	-0.16
TandEC	**AbsMeanDiff**	66.2	22.1	1
	Sd	74.6	28.1	0.8
	ICC	0.43	0.02	0.03
	SEM	56.5	27.9	0.78
	%MDC	7.2	58.5	50.7
	Effect size	-0.20	-0.16	-0.46
SLSEO	**AbsMeanDiff**	27.1	5.1	1.5
	Sd	13.8	6.4	1.1
	ICC	0.80	0.69	0.15
	SEM	6.2	3.5	1.0
	%MDC	2.8	28.8	57.7
	Effect size	-0.02	0.23	0.35
SLSEC	**AbsMeanDiff**	201.4	117.9	2.6
	Sd	286.9	108.1	3.7
	ICC	0.77	0.95	0.71
	SEM	137.2	25.2	2.0
	%MDC	5.3	12.9	49.2
	Effect size	0.08	0.11	0.42

DLSFNEO, double leg stance feet natural eyes open; DLSFNEC, double leg stance feet natural eyes closed; DLSFTEO, double leg stance feet together eyes open; DLSFTEC, double leg stance feet together eyes closed; TandEO, Tandem stance eyes open; TandEC, Tandem stance eyes closed; SLSEO, Single leg stance eyes open; SLSEC, Single leg stance eyes closed; PL, path length; RMS, root mean squared; AbsMeanDiff, absolute mean difference between day one and day two; ICC, intraclass correlation coefficient; SEM, standard error of measurement; MDC, minimal detectable change; Sd, standard deviation.

Balance Test Report

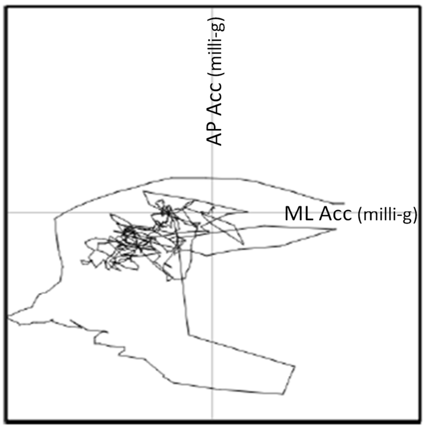
Name:

Date:

Eyes:

Task Time:

	AP	ML	Total	
PL	185.9	105.3	291.2	mg
JERK	45.71	20.35	66.06	mg^2/s
RMS	2.697	2.099	3.418	mg



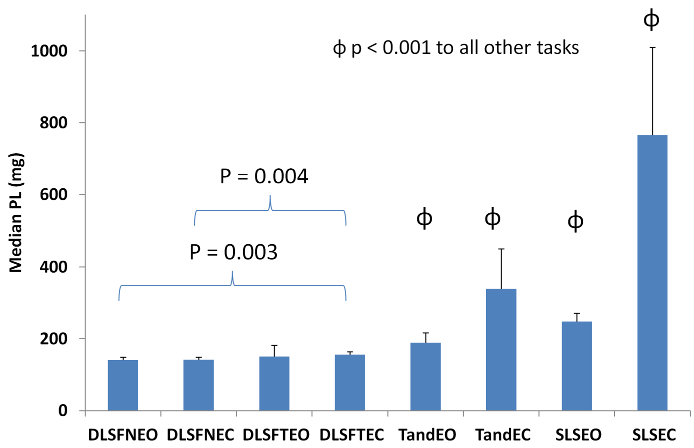

ϕ p < 0.001 to all other tasks
ϕ
ϕ
ϕ
ϕ
P = 0.004
P = 0.003
Median PL (mg)
1000
800
600
400
200
0
DLSFNEO
DLSFNEC
DLSFTEO
DLSFTEC
TandEO
TandEC
SLSEO
SLSEC

Figure 2a

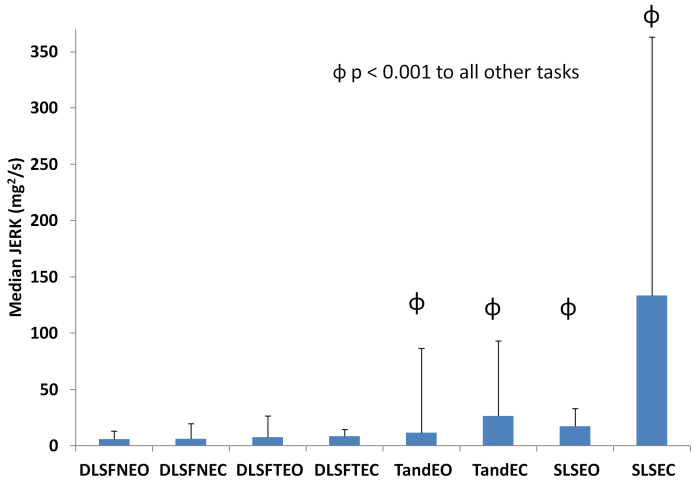

ϕ p < 0.001 to all other tasks
Median JERK (mg²/s)
350
300
250
200
150
100
50
0
ϕ
ϕ
ϕ
ϕ
DLSFNEO
DLSFNEC
DLSFTEO
DLSFTEC
TandEO
TandEC
SLSEO
SLSEC

Figure 2b

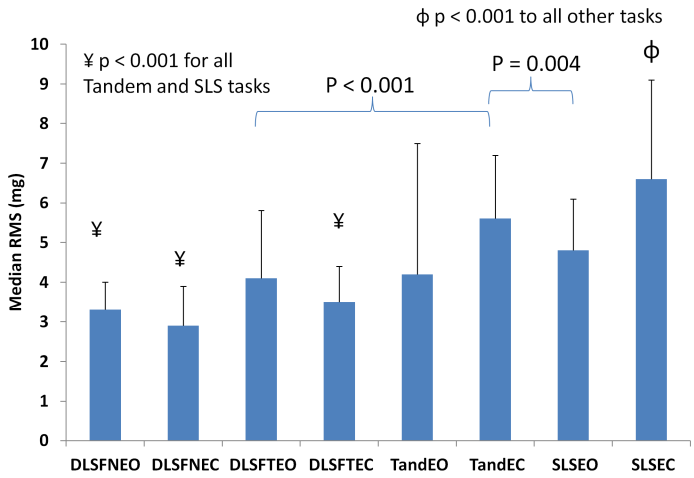

ϕ p < 0.001 to all other tasks
¥ p < 0.001 for all Tandem and SLS tasks
P < 0.001
P = 0.004
ϕ
¥
¥
¥
Median RMS (mg)
0
1
2
3
4
5
6
7
8
9
10
DLSFNEO
DLSFNEC
DLSFTEO
DLSFTEC
TandEO
TandEC
SLSEO
SLSEC

Figure 2c

Highlights

Balance performance can be reliably measured using a sacral mounted balance sensor.

This study recommends quantification of balance using path length (PL).

Findings show changes in PL of more than 10% represents true change in performance.