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The Quantification of Hop Landing Balance Using Trunk-Mounted Accelerometry

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Context: Balance is important for injury prediction, prevention, and rehabilitation. Clinical measurement of higher level balance function such as hop landing is necessary. Currently, no method exists to quantify balance performance following hopping in the clinic. ***Objective:*** To quantify the sacral acceleration profile and test–retest reliability during hop landing. ***Participants:*** A total of 17 university undergraduates (age 27.6 [5.7] y, height 1.73 [0.11] m, weight 74.1 [13.9] kg). ***Main Outcome Measure:*** A trunk-mounted accelerometer captured the acceleration profile following landing from hopping forward, medially, and laterally. The path length of the acceleration traces were computed to quantify balance following landing. ***Results:*** Moderate to excellent reliability (intraclass correlation coefficient .67–.93) for hop landing was established with low to moderate SEM (4%–16%) and minimal detectable change values (13%–44%) for each of the hop directions. Significant differences were determined in balance following hop landing from the different directions. ***Conclusion:*** The results suggest that hop landing balance can be quantified by trunk-mounted accelerometry.

Keywords: accelerometer, sway, path length

Balance is important for injury prediction, rehabilitation,[1] and prevention.[2] However, clinic-based balance measurement is often constrained to subjective judgment or task duration. This fails to determine the quality of balance performance and lacks detailed objectivity necessary for quantifying subtle changes. Hop testing is highly prevalent in lower limb rehabilitation, especially post knee surgery or in patellofemoral pain. Measuring quality of landing is challenging for clinicians using hop testing. Laboratory-based systems that quantify balance often require specific fixed environments and incur increased costs, limiting their uptake into routine practice. Therefore, novel methods for quantifying balance in clinical practice are needed.

Accelerometers have quantified balance across a range of disease states and task conditions.[3–6] Accelerometers commonly mounted on the low back or sacral area have high reported correlation with force plate measures of balance.[3,7,8] Although it is acknowledged that the 2 measurement techniques measure subtly different constructs of balance (sway of center of pressure vs sacral acceleration), their relationship suggests that accelerometry offers a valid measurement method for balance. Furthermore, the reliability of such methods is high across a range of tasks from double-leg, single-leg to tandem stance.[6] Despite this, highly dynamic balance tasks such as hop landing have yet to be investigated. Testing single-leg hop landing is important in the assessment of return to sport readiness and as such clinicians draw on the between-leg comparisons for this assessment. However, in the absence of reliable quantification, between-leg differences remain in the domain of estimation.

The aim of this study was to use an accelerometer to quantify the sacral acceleration profile and test–retest reliability during hop landing.

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Methods

Participants

A total of 17 participants (mean [SD]: age 27.6 [5.7] y, height 1.731 [0.105] m, mass 74.1 [13.9] kg) were recruited through social media advertisement. This was based on a sample size calculation using the method outlined in Walter et al[9] with $\alpha = .05$, $\beta = 80\%$, 3 repetitions of the task, and desirable and minimal correlation values set at .8 and .5, respectively. Participants were excluded if they reported any previous injury requiring plaster cast or surgery, current or previous injury to either lower limb in the preceding 12 months, current or previous head injury/concussion, current or previous known neurological disorder, known balance issues (eg, vertigo), or aged >50 years. The study was approved by the Bournemouth University Research Ethics Committee, and all participants provided written informed consent prior to taking part.

Experimental Protocol

All data were collected within a clinical skills suite at the university. Participants' age, height, and weight along with dominant limb were recorded.[10] The sensor was fixed to the skin over the L4-S1 spinous processes using medical grade double-sided tape. The balance sensor (THETAmetrix, Hampshire, United Kingdom) houses a triaxial accelerometer and gyroscope, which communicates wirelessly to a PC, and data were captured at 100 Hz. Previously, similar methodology (attachment, sensing elements, software, etc) has been used to assess balance during various double- and single-leg balance tasks demonstrating good to excellent reliability (intraclass correlation coefficient, ICC > .7).[6]

Landing Task

Participants performed single-leg hopping with a "fixed" landing in the forward, lateral, and medial directions. Floor markers were used to denote start and landing positions. Hop distances were normalized to 50% the individual's height (forward hopping) and 33% the individual's height (medial and lateral hopping). The hop was deemed successful if the participant landed with their foot touching the floor marker and balance maintained for >2 seconds. One practice attempt was permitted prior to 3 hop landings being captured. The order of hopping was standardized to dominant prior to nondominant and forward hopping followed by medial and lateral hopping.

Data Processing

Data were transferred to MATLAB 2008b (The MathWorks, Inc, Natick, MA) for processing. Raw accelerations were used to calculate the resultant acceleration vector by taking the square root of the sum of squared accelerations for each axis. The impact peak was identified, denoting the landing and its time index recorded. The acceleration data were corrected for tilt using the angle derived from the gyroscopes and used to correct for sensor tilt at each time point, removing the gravity vector and thus sensor data represented true anteroposterior and mediolateral accelerations. These accelerations were trimmed from the landing index to 1 second following landing, and the mediolateral and anteroposterior accelerations were then plotted to produce a postural sway plot.

Table 1 Hop Landing Path Length for Dominant and Nondominant

Path length, mg	Dominant (median [IQR])	Nondominant (median [IQR])
Hop forward	15.9 (3.5)	18.1 (6.0)
Hop lateral	13.7 (5.2)	15.3 (4.4)
Hop medial	15.9 (2.6)	14.4 (3.5)

Abbreviations: IQR, interquartile range; mg, milligravity.

The path length of this sway trace was calculated from the sum of the difference between 2 sequential data points (sample $(x+1) -$ sample (x)). Therefore, to quantify hop landing postural sway, the path length of the mediolateral and anteroposterior accelerations were summed for 1 second following hop landing. Three trials were used to quantify ICC and the mean of 3 trials used for between-condition comparison.

Statistical Analysis

Statistics were processed using MATLAB and RealStats in Excel. Reliability was explored using $ICC_{3,k}$. In order to understand the natural variability of such tasks, the SEM was calculated along with the minimal detectable change ($MDC_{95\%}$) using the following equation:

$$MDC_{95\%} = 1.96 \times SEM \times \sqrt{2}.$$

In addition, the task complexity was explored using Kruskal–Wallis and post hoc Mann–Whitney U tests as data were not normally distributed.

Results

The mean (SD) for the average of 3 hop landing trials are presented in Table 1, and a typical sway trace is presented in Figure 1A and 1B.

Hop landing ICC ranged from .673 to .929 indicating excellent to moderate reliability (Table 2). Dominant and nondominant medial hop landing had the greatest ICC, suggesting greater consistency in landing balance. Hop forward had the largest SEM and this remained the case once normalized, suggesting greater variability in hop landing performance. This subsequently influenced the MDC values which suggest that with 95% confidence, a change of 7 mg or 37% in path length for hop forward

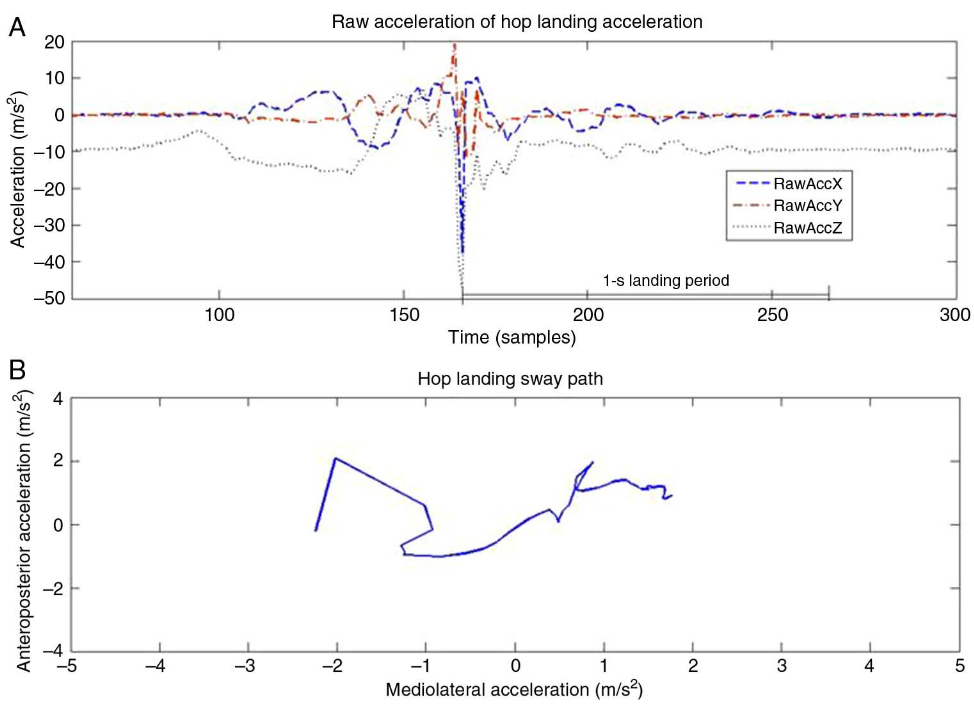

Figure 1 — Raw acceleration for hopping task (A) and mediolateral against anteroposterior acceleration sway for 1s following landing (B).

Table 2 Reliability and Variability Estimates From Hop Landing

Task	ICC	SEM, mg	SEM as % of task median	$MDC_{95\%}$	MDC as % of task median
D forward	.673	2.5	15.7	7.0	44.0
D lateral	.770	1.5	10.9	4.2	30.7
D medial	.929	0.7	4.4	2.0	12.6
ND forward	.753	2.2	12.2	6.2	34.3
ND lateral	.702	1.4	9.2	4.0	26.1
ND medial	.842	1.3	9.0	3.5	24.3

Abbreviations: D, dominant; ICC, intraclass correlation coefficient; MDC, minimal detectable change; mg, milligravity; ND, nondominant.

landing represents true change. Landing from hopping in other directions had lower MDCs around 25% and overall the average MDC for all hops was 26.4%.

Kruskal–Wallis testing demonstrated a significant difference across the tasks ($\chi^2_5 = 12.81$, $P = .03$). There were significant differences found between dominant hop forward versus nondominant medial hop landing ($P = .03$), dominant hop lateral versus nondominant hop forward landing ($P = .04$), and nondominant hop forward versus nondominant hop medial landing ($P = .05$).

Discussion

This study set out to determine whether trunk-mounted accelerometry could quantify postural sway during hop landing. Previous studies have demonstrated the use of trunk-mounted accelerometry for measuring balance. The findings of this study suggest that accelerometry can be used to quantify balance during hop landing.

Previous studies investigating trunk-mounted accelerometry have demonstrated similar ICC values for single-leg stance and tandem stance (.69–.89).[3,6] Therefore, despite the highly dynamic nature of hop landing, the reliability values are consistent with less ballistic tasks reported in the literature. Reporting the variability of repeat performance in the form of SEM is important. The results demonstrate that SEMs < 10% for the medial and lateral hop landings and slightly more (<14%) for forward hop landing, which demonstrates a high degree of consistency for this task, similar to that in the literature (6%–32%).[3,6] The MDC offers a confidence level for the ability to detect true change beyond natural variability of the specific task. This study demonstrated that with 95% confidence, a change >27% is likely to represent a true change in performance in landing balance from medial or lateral hops. The MDC is affected by the variability of repeat performance and as such mirrors the findings of the SEM. MDC values have not been widely reported in the literature, but MDC values of 13% to 91% for tasks ranging from double-leg, single-leg to tandem stance have been noted.[6] Therefore, the findings suggest that hop landing balance as measured by trunk accelerometry is similar to other tasks in its ability to detect true change in performance.

Landing from forward hopping was more variable. This may reflect the additional task demand of hopping further (50% height) compared with the other directions (33% height). Hopping further would result in greater force to arrest the motion and greater levels of coordination for balance to be maintained. Indeed, it was demonstrated that this task had path lengths around 10% to 20% greater than the other directions.

It is easy to assume that the results of this study solely represent measurement variability, that is, the device. However, the results reflect the human and device interaction and the variability of this "coupling." Some error and thus variability will lie with the sensor. Bench top experiments and calibration procedures identify this error to be typically <1%. Another source of error is the sensor–human interface notably the soft-tissue motion created by the landing from a ballistic task. The device was securely fastened, but no additional external reinforcement was used. Previous studies have demonstrated little impact from the skin motion artifact for acceleration signals;[11] however, they did not investigate hopping. Finally, there will be human error. The human during completion of such a task will have natural variation in performance. Some highly practiced movements are very consistent; however, in the current study, individuals were not "highly practiced in hopping." Therefore, some of the variations in measures reported in this study are likely to be due to task-specific movement inconsistencies.

This study demonstrates that trunk-mounted accelerometry can be used to quantify hop landing balance. Reliability of repeated hop landing measurements was good, and SEM and MDC values suggest that such a method can be used within the clinical setting. Trunk-mounted accelerometry should be considered by clinicians and researchers wanting to quantify hop landing balance.

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