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The quantification of hop landing balance using trunk mounted accelerometry.

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Abstract

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: The objective of this study was to quantify the sacral acceleration profile and test-retest reliability during hop landing.

Participants: Seventeen university undergraduates (Age 27.6(5.7) years, Height 1.73(0.11) m, Weight 74.1(13.9)kg).

Outcome Measure: A trunk mounted accelerometer captured the acceleration profile following landing from hopping forwards, medially and laterally. The path length of the acceleration traces were computed to quantify balance following landing.

Results: Moderate-to-excellent reliability (ICC 0.67-0.93) for hop landing was established with low-to-moderate standard error of measurement (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 hop landing balance can be quantified by trunk mounted accelerometry.

Introduction

Balance is important for injury prediction, rehabilitation (Emery et al., 2005) and prevention (Steffen et al., 2013). However clinic based balance measurement is often constrained to subjective judgement 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 which 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 (Mancini et al., 2012; Marchetti et al., 2013; Saunders et al., 2015; Williams et al., 2016). Accelerometers commonly mounted on the low back or sacral area have high reported correlation with force-plate measures of balance (Adlerton et al., 2003; Mancini et al., 2012; Whitney et al., 2011). Although it is acknowledged that the two measurement techniques measure subtly different constructs of balance (sway of COP vs sacral acceleration), their relationship suggests 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 and tandem stance (Williams et al., 2016). 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 remains to domain of estimation.

Human Kinetics, 1607 N Market St, Champaign, IL 61825

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

Methods

Participants

Seventeen participants (mean (standard deviation (sd)); Age 27.6(5.7) years, 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. (1998) with alpha =0.05, Beta=80%, 3 repetitions of the task and desirable and minimal correlation values set at 0.8 and 0.5 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 (e.g. 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 university. Participants' age, height and weight along with dominant limb were recorded (Hoffman et al., 1998). The sensor was fixed to the skin over the L4-S1 spinous processes using medical grade double sided tape. The balance sensor (THETAmetrix, Hampshire, UK) houses a triaxial accelerometer and a gyroscope which communicates wirelessly to a PC and data were captured at 100Hz. 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 (ICC>0.7) (Williams et al., 2016).

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 normalised 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 three hop landings being captured. The order of hopping was standardised to dominant prior to non-dominant and forward hopping followed by medial and lateral hopping.

Data processing

Data were transferred to MatLab (Mathworks, 2008b) 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 anterioposterior and mediolateral accelerations. These accelerations were trimmed from the landing index to 1 second following landing and the mediolateral and anterioposterior accelerations were then plotted to produce a postural sway plot. 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 anteriorposterior accelerations were summed for 1 second following hop landing. Three trials were used to quantify ICC and the mean of three trials used for between condition comparison.

Statistical analysis

Statistics were processed using MatLab and RealStats in Excel. Reliability was explored using Intraclass Correlation Coefficients ($ICC_{3,k}$). In order to understand the natural variability of such tasks the Standard Error of Measurement (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 three hop landing trials are presented in table 1 and a typical sway trace is presented in figure 1a and b.

Hop landing ICC ranged from 0.673-0.929 indicating excellent-to-moderate reliability (table 2). Dominant and non-dominant 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 normalised, suggesting greater variability in hop landing performance. This subsequently influenced the MDC values which suggest that with 95% confidence a change of 7mg or 37% in path length for hop forward 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, X^2 (5) = 12.81, p = 0.025.

There were significant differences between dominant forward vs non-dominant medial hop landing (p = 0.0341).

There were significant differences between dominant hop lateral vs non-dominant hop forward landing (p = 0.0424).

There were also significant differences between non-dominant hop forward vs non-dominant hop medial landing (p = 0.0466).

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 (SLS) and tandem stance (0.69-0.89) (Mancini et al., 2012; Williams et al., 2016). 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 SEMs < 10% for the medial and lateral hop landings and slightly more (<14%) for forward hop landing demonstrating a high degree of consistency for this task, similar to that in the literature (6%-32%) (Mancini et al., 2012; Williams et al., 2016). 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%-91% for tasks ranging from double leg, single leg to tandem stance have been noted (Williams et al., 2016). 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 to the other directions (33% height). Hopping further would result in greater force to arrest the motion and greater levels of co-ordination for balance to be maintained. Indeed it was demonstrated that this task had path lengths around 10%-20% greater than the other directions.

It is easy to assume that the results of this study solely represent measurement variability i.e. 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 artefact for acceleration signals (Morgado Ramirez et al., 2013) 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 practised in hopping'. Therefore some of the variation 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 such a method is usable within the clinical setting. Trunk mounted accelerometry should be considered by clinicians and researchers wanting to quantify hop landing balance.

Conflict of Interest Statement

[redacted].

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Table 1. Hop landing path length for Dominant and Non-Dominant

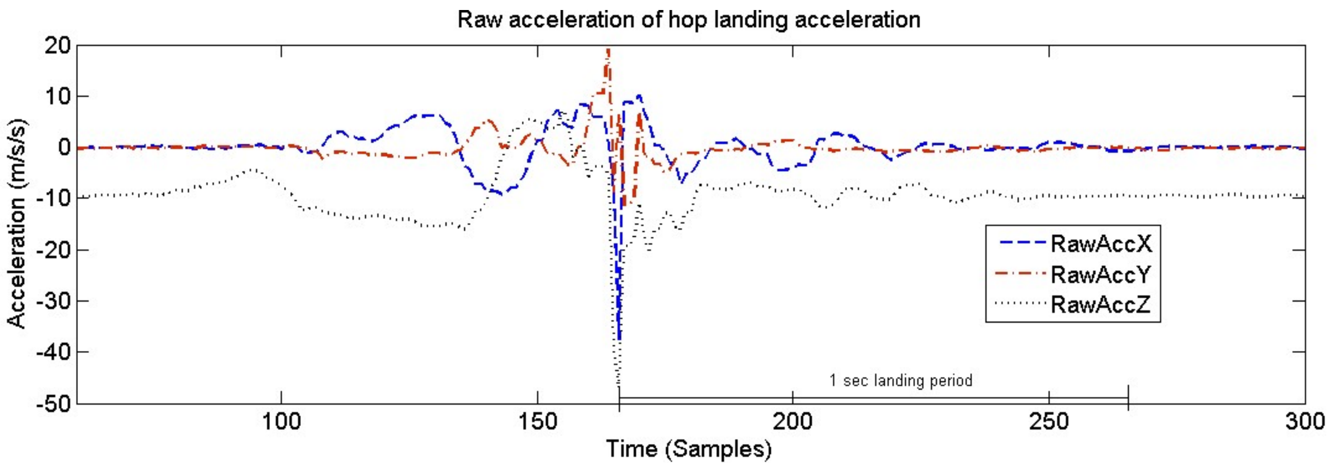
Path length (mg)	Dominant (Median (IQR)	Non-Dominant (Median (IQR)
Hop Forward	15.9 (3.5)	18.1 (6.0)
Hop Lateral	13.7 (5.2)	15.3 (4.4)
Hop Medial	15.9 (2.6)	14.4 (3.5)

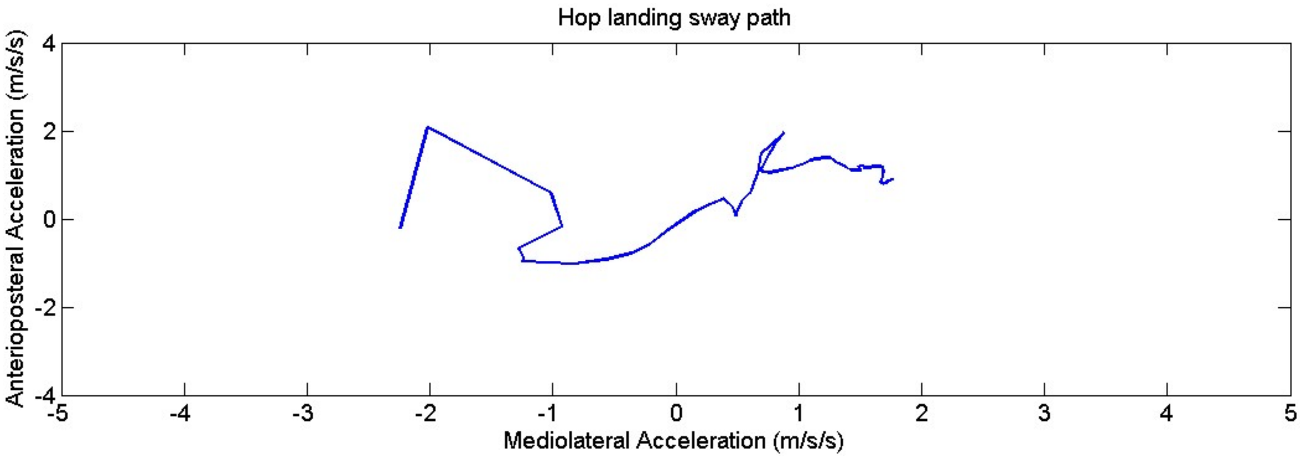
mg; milli-gravity, IQR; inter-quartile range.

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	0.673	2.5	15.7	7.0	44.0
D lateral	0.770	1.5	10.9	4.2	30.7
D medial	0.929	0.7	4.4	2.0	12.6
ND forward	0.753	2.2	12.2	6.2	34.3
ND lateral	0.702	1.4	9.2	4.0	26.1
ND medial	0.842	1.3	9.0	3.5	24.3

D; dominant, ND; non-dominant, ICC; intra-class correlation coefficient, SEM; standard error of measurement, MDC; minimal detectable change.


Raw acceleration of hop landing acceleration
Acceleration (m/s/s)
20
10
0
-10
-20
-30
-40
-50
100
150
200
250
300
Time (Samples)
RawAccX
RawAccY
RawAccZ
1 sec landing period


Hop landing sway path
Anterioposteral Acceleration (m/s/s)
4
2
0
-2
-4
-5
-4
-3
-2
-1
0
1
2
3
4
5
Mediolateral Acceleration (m/s/s)