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An exploration of segment acceleration and angular velocity during different balance conditions measures in the assessment of stability

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Abstract

Objectives

To identify the most sensitive measures to assess postural stability and to detect between leg differences.

Methods

Seventeen healthy volunteers participated in this University physiotherapy clinic experiment. Linear segment acceleration and segment angular velocity data were collected using Inertial Measurement Units on the left and right calcanei, shanks, and pelvis. Participants performed a 30s single-leg stance on a firm surface and a balance pad on their dominant and non-dominant leg. A two factor (2x2) repeated measures analysis of variance was used to compare the two balance surfaces and the leg dominance.

Results

Comparing the surfaces, both highest effect size and percentage difference were found in the pelvis angular velocity data. Apart from the anteroposterior linear acceleration of the shank, both effect size and percentage change showed that all average linear acceleration values were less sensitive to detect change between surfaces when compared to the average angular velocity values of the same body segments.

Conclusions

The angular velocity appeared to be more sensitive than the acceleration when differentiating between the two levels of stability challenge. Inertial measurement units' measures have the potential to be used in the evaluation of postural stability and balance strategies within clinical assessment.

Keywords:

postural balance; inertial measurement unit; single-leg stance; clinical assessment

INTRODUCTION

Most lower limb sports-related musculoskeletal injuries affect the postural stability and the joint stability (1). This loss of stability can in turn have a negative impact on the quality of life and the risk of re-injury after the return to sport (2,3). Since the assessment of stability might have the potential to decrease the risk of re-injury, or avoid sub-optimal treatment outcomes, it is of great interest to establish and optimize its diagnostic efficacy (1). Various methods have been explored to enable more accurate objective monitoring of the improvement in the affected limb (1,4), which has included the use of software, sensors, and robotics (1,4).

Contrarily to functional and physiological assessments, quantitative assessment of balance using force plates, pressure plates and inertial measurement units have been shown to avoid evaluator bias (5–7). In addition, such instrumented balance measures also appear to have good face validity and sensitivity when assessing balance, and have been shown to offer important outcome measures in clinical practice (5,8).

IMUs sensors, which can measure linear and angular motion in 3D space, have the potential to substitute the force platform measures during quiet standing tasks (5,9). In addition, they have the advantage of being small, lightweight, inexpensive, robust with a long battery life enabling data collection within the real-world environment. These can be attached to any part of the body, and often have on board data storage or the ability to data transfer data wirelessly (10).

The evaluation of postural stability with IMUs is based on postural sway measurements (11). Specific measures using accelerometry of the pelvis have been demonstrated to be more sensitive than force platform measurements in detecting differences during balance challenging tasks (12). Moreover, the mediolateral, anteroposterior and resultant root mean square acceleration of IMUs located on the projection of center of mass have been shown to be able to distinguish different balance challenging tasks (8,9,13,14). However, these measurements considered the

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linear acceleration of the center of mass of the whole body rather than the control adjustments of individual body segments or side to side differences.

The postural differences between injured and uninjured leg can enable healthcare professionals to make informed decisions regarding the return to sport. While this difference might be highlighted via IMUs linear acceleration measurements during Single Limb Stance (SLS), little or no published data exists regarding the use of IMUs angular velocity measurements during SLS (1).

This current study explores the assessment of the calcaneus, shank and pelvis acceleration and angular velocity, to determine their sensitivity in the assessment of balance. Moreover, the use of linear segment acceleration and segment angular velocity may give a greater understanding of possible strategies for maintaining postural stability. It was hypothesized that angular velocity may offer a more sensitive measure to detect different balance conditions and between leg differences than linear acceleration. By measuring the acceleration and angular velocity of three body locations, this study also explored which sensor location offered the most sensitive measures to assess postural stability and to detect between leg differences.

METHODS

Participants: The inclusion criteria were; aged between 18 and 65 years, with no musculoskeletal injury or history of surgery to the lower extremities within the previous 6 months, and no neurologic disorders. Seventeen individuals (9 males and 8 females) were recruited from a staff and student population. All participants read the information sheet for the study and signed the consent form approved by the University Ethics Committee prior to participation.

Experimental protocol: Segment acceleration and angular velocity data were collected using Delsys Trigno IM sensors (Trigno™ System, Delsys Inc., USA), which measured acceleration and angular velocity using accelerometers and gyroscopes at 148Hz. The sensors were attached to the skin with double-sided tape on the posterior aspect calcaneus (calcaneus), 2 centimetres above the lateral malleolus on the left and right legs (shank) and between the postero-superior iliac spines (pelvis) (figure 1).

Figure 1: Calcaneus, shank and pelvis IMUs locations



After attachment of the sensors, the participants were asked to perform a barefoot single-leg stance on a firm surface task (SLS-firm) and a single-leg stance on a foam pad (SLS-foam), in a randomised order. Each of these tasks were performed first on the dominant leg and then on the non-dominant leg. In order to avoid physical fatigue, no practice trials were performed. Limb dominance was assessed by identifying the preferred leg to kick a ball. The subjects were asked to stand as still as possible for 30 seconds with the hands placed on the hips and the non-weight bearing foot near the knee of the stance leg without touching it. Participants carried out these tasks once on each foot under each condition. The firm surface was a concrete floor with a thin carpet layer and the foam surface was an Airex Balance-pad (Airex AG, Switzerland) with the dimensions of 50 cm x 41 cm x 6 cm. If participants put their non-supportive foot on the ground during a task they were asked to perform the task again, if they were unable to complete the task then a missing value was entered. This occurred on two occasions with two different participants.

Static postural stability parameters: Segment acceleration and angular velocity data, using the sensor coordinate systems, were exported from EMGworks (Delsys Inc.) into c3d format for further data processing and analysis in Visual 3D (C-motion Inc, USA). The root mean square (RMS) relative to the mean and to the maximum from the linear acceleration and angular velocity signals were calculated from the middle 10 seconds of each trial as followed: $RMS = \sqrt{1/n\ (x_1^2 + x_2^2 + \ldots + x_n^2)}$

Statistical analysis: A two factor (2x2) repeated measures analysis of variance was used to compare the two balance surfaces and the leg dominance and to determine any interactions between surface and leg dominance for the linear acceleration and angular velocity data for the different axes for the different body segments. Post hoc pairwise comparisons with Least Significant Difference were performed between the two tasks and of the leg dominance. The effect size was calculated using partial eta squared which was used for evaluating the magnitudes of the observed differences. An alpha level of 0.05 was set a priori. All the statistical analyses were carried out in SPSS version 23 (IBM, Armonk, USA). In addition, the percentage of mean difference between the SLS-firm and SLS-foam RMS data was calculated as follows:

$$\frac{(\text{RMS SLS foam} - \text{RMS SLS-firm}) \times 100}{\text{RMS SLS-firm}}$$

Finally, the percentage of mean difference between the dominant leg and the non-dominant leg RMS data was calculated as follows:

$$\frac{(\text{RMS Non-dominant} - \text{RMS Dominant}) \times 100}{\text{RMS Dominant}}$$

Table 1: Mean differences in surface and side to side pelvic estimates

Variable	Average Root Mean Square values					Maximum Root Mean Square values				
	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences
Surface differences										
Medio-lateral acceleration	0.150 (0.018)	0.150 (0.018)	0.0	*p*=0.992 ηp²=0.000	-0.050 to 0.050	0.183 (0.019)	0.290 (0.032)	58.5	p=0.010* ηp²=0.37	-0.184 to -0.030
Proximal-Distal acceleration	1.113 (0.008)	1.099 (0.009)	-1.3	*p*=0.006* ηp²=0.402	0.005 to 0.023	1.147 (0.010)	1.178 (0.011)	2.7	p<0.001* ηp²=0.593	-0.045 to -0.017
Antero-poster. acceleration	0.230 (0.035)	0.279 (0.034)	21.3	*p*=0.003* ηp²=0.462	-0.077 to -0.020	0.262 (0.034)	0.377 (0.041)	43.9	p=0.002* ηp²=0.48	-0.180 to -0.049
Sagittal angular velocity	1.57 (0.05)	2.98 (0.36)	89.5	*p*=0.001* ηp²=0.543	-2.12 to -0.70	5.613 (0.656)	14.96 (3.21)	166.6	p=0.004* ηp²=0.443	-15.12 to -3.58
Transverse angular velocity	1.82 (0.23)	5.33 (0.58)	192.3	*p*<0.001* ηp²=0.826	-4.39 to -2.62	8.193 (1.443)	31.14 (5.51)	280.1	p<0.001* ηp²=0.644	-32.34 to -13.55
Coronal angular velocity	1.32 (0.07)	4.41 (0.82)	233.3	*p*=0.001* ηp²=0.514	-4.74 to -1.43	4.87 (0.34)	22.59 (4.54)	363.5	p=0.001* ηp²=0.524	-27.00 to -8.43
Side to side differences										
Medio-lateral acceleration	0.200 (0.025)	0.100 (0.018)	-50.0	*p*=0.010* ηp²=0.371	-0.172 to -0.028	0.302 (0.035)	0.172 (0.020)	-43.0	*p*=0.008* ηp²=0.387	-0.220 to -0.040
Proximal-Distal acceleration	1.104 (0.009)	1.108 (0.007)	0.4	*p*=0.375 ηp²=0.053	-0.006 to 0.015	1.166 (0.013)	1.158 (0.008)	-0.7	*p*=0.352 ηp²=0.058	-0.027 to 0.010
Antero-poster. acceleration	0.258 (0.035)	0.251 (0.033)	-2.7	*p*=0.547 ηp²=0.025	-0.028 to 0.016	0.333 (0.037)	0.305 (0.034)	-8.4	*p*=0.161 ηp²=0.127	-0.069 to 0.013
Sagittal angular velocity	2.44 (0.24)	2.11 (0.19)	-13.2	*p*=0.096 ηp²=0.174	-0.71 to 0.06	11.64 (2.20)	8.94 (1.75)	-23.1	*p*=0.054 ηp²=0.225	-5.45 to 0.06
Transverse angular velocity	3.75 (0.41)	3.41 (0.44)	-9.2	*p*=0.347 ηp²=0.059	-1.10 to 0.41	21.46 (3.73)	17.87 (3.51)	-16.7	*p*=0.193 ηp²=0.11	-9.21 to 2.03
Coronal angular velocity	3.14 (0.48)	2.59 (0.43)	-17.3	*p*=0.095 ηp²=0.175	-1.19 to 0.11	16.66 (3.18)	10.80 (2.18)	-35.2	*p*=0.047* ηp²=0.239	-11.63 to -0.10

Firm: Firm surface; Foam: Foam surface; % Diff: Percentage difference; 95% CI: 95% confidence interval; Dom: dominant; *: significant difference

Table 1: Mean differences in surface and side to side calcaneal estimates

Variable	Average Root Mean Square values					Maximum Root Mean Square values				
	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences
Surface differences										
Medio-lateral acceleration	0.072 (0.010)	0.113 (0.012)	56.9	p=0.018* ηp^2=0.319	-0.074 to -0.008	0.143 (0.017)	0.416 (0.042)	190.9	p<0.001* ηp^2=0.697	-0.371 to -0.173
Proximal-Distal acceleration	0.971 (0.016)	0.934 (0.028)	-3.8	p=0.279 ηp^2=0.078	-0.034 to 0.109	0.996 (0.012)	1.102 (0.014)	10.6	p<0.001* ηp^2=0.768	-0.138 to -0.074
Antero-poster. acceleration	0.310 (0.030)	0.364 (0.030)	17.4	p=0.146 ηp^2=0.136	-0.130 to 0.021	0.343 (0.040)	0.406 (0.021)	18.4	p=0.092 ηp^2=0.178	-0.137 to 0.011
Sagittal angular velocity	2.90 (0.04)	4.73 (0.48)	62.9	p=0.002* ηp^2=0.49	-2.85 to -0.80	6.40 (0.38)	21.84 (3.69)	241.4	p=0.001* ηp^2=0.544	-23.20 to -7.70
Transverse angular velocity	1.86 (0.16)	5.08 (0.55)	173.5	p<0.001* ηp^2=0.695	-4.40 to -2.05	9.0 (0.72)	30.70 (4.19)	241.1	p<0.001* ηp^2=0.655	-30.30 to -13.00
Coronal angular velocity	2.95 (0.23)	11.74 (1.17)	298.3	p<0.001* ηp^2=0.817	-11.10 to -6.50	14.71 (1.65)	62.20 (7.50)	322.6	p<0.001* ηp^2=0.776	-61.50 to -33.40
Side to side differences										
Medio-lateral acceleration	0.088 (0.009)	0.097 (0.013)	10.2	p=0.603 ηp^2=0.018	-0.027 to 0.044	0.309 (0.037)	0.250 (0.017)	-19.1	p=0.132 ηp^2=0.145	-0.140 to 0.020
Proximal-Distal acceleration	1.008 (0.009)	0.897 (0.026)	-11.0	p<0.001* ηp^2=0.589	-0.163 to -0.060	1.092 (0.016)	1.007 (0.020)	-7.8	p=0.008* ηp^2=0.381	-0.146 to -0.025
Antero-poster. acceleration	0.291 (0.020)	0.383 (0.036)	31.6	p=0.011* ηp^2=0.358	0.024 to 0.159	0.348 (0.021)	0.401 (0.039)	15.2	p=0.122 ηp^2=0.152	-0.016 to 0.123
Sagittal angular velocity	3.49 (0.36)	4.14 (0.23)	18.6	p=0.095 ηp^2=0.175	-0.13 to 1.42	14.583 (2.887)	13.653 (1.380)	-6.4	p=0.715 ηp^2=0.009	-6.26 to 4.40
Transverse angular velocity	3.45 (0.38)	3.48 (0.27)	0.9	p=0.917 ηp^2=0.001	-0.60 to 0.66	21.190 (3.338)	18.495 (1.759)	-12.7	p=0.377 ηp^2=0.052	-9.01 to 3.62
Coronal angular velocity	7.54 (0.77)	7.15 (0.64)	-5.2	p=0.487 ηp^2=0.033	-1.56 to 0.78	42.01 (5.54)	34.88 (4.61)	-17.0	p=0.208 ηp^2=0.103	-18.67 to 4.43

Firm: Firm surface; Foam: Foam surface; % Diff: Percentage difference; 95% CI: 95% confidence interval; Dom: dominant; *: significant difference

Table 3: Mean differences in surface and side to side estimates obtained from the shank

Variable	Average Root Mean Square values					Maximum Root Mean Square values				
	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences	Firm	Foam	% Diff	*p* value Effect size	95% CI of the Differences
Surface differences										
Medio-lateral acceleration	0.216 (0.015)	0.211 (0.013)	-2.3	p=0.475 ηp^2=0.035	-0.009 to 0.019	0.288 (0.015)	0.541 (0.037)	87.8	p<0.001* ηp^2=0.779	-0.327 to -0.179
Proximal-Distal acceleration	0.996 (0.005)	0.969 (0.023)	-2.7	p=0.257 ηp^2=0.085	-0.022 to 0.076	1.043 (0.004)	1.125 (0.014)	7.9	p<0.001* ηp^2=0.723	-0.110 to -0.054
Antero-poster. acceleration	0.125 (0.017)	0.165 (0.029)	32.0	p=0.240 ηp^2=0.091	-0.110 to 0.030	0.197 (0.025)	0.312 (0.018)	58.4	p=0.005* ηp^2=0.421	-0.189 to -0.041
Sagittal angular velocity	2.040 (0.121)	3.919 (0.452)	92.1	p<0.001* ηp^2=0.601	-2.722 to -1.037	8.591 (1.053)	18.765 (2.736)	118.4	p=0.001* ηp^2=0.544	-15.304 to -5.044
Transverse angular velocity	8.220 (0.698)	14.016 (1.108)	70.5	p<0.001* ηp^2=0.737	-7.702 to -3.890	36.411 (4.196)	71.459 (6.340)	96.3	p<0.001* ηp^2=0.717	-47.16 to -22.94
Coronal angular velocity	5.558 (0.156)	6.578 (0.470)	18.3	p=0.050 ηp^2=0.233	-2.038 to -0.002	17.207 (1.286)	27.760 (3.816)	61.3	p=0.010* ηp^2=0.367	-18.17 to -2.93
Side to side differences										
Medio-lateral acceleration	0.203 (0.015)	0.224 (0.015)	10.3	p=0.127 ηp^2=0.148	-0.007 to 0.049	0.421 (0.031)	0.409 (0.020)	-2.8	p=0.680 ηp^2=0.012	-0.071 to 0.048
Proximal-Distal acceleration	1.039 (0.014)	0.927 (0.014)	-10.8	p<0.001* ηp^2=0.781	-0.145 to -0.080	1.127 (0.017)	1.041 (0.014)	-7.6	p=0.006* ηp^2=0.405	-0.143 to -0.028
Antero-poster. acceleration	0.152 (0.019)	0.137 (0.032)	-9.9	p=0.710 ηp^2=0.009	-0.099 to 0.069	0.291 (0.015)	0.219 (0.027)	-24.7	p=0.057 ηp^2=0.221	-0.145 to 0.002
Sagittal angular velocity	3.01 (0.35)	2.95 (0.23)	-2.1	p=0.826 ηp^2=0.003	-0.66 to 0.53	14.60 (2.29)	12.76 (1.74)	-12.6	p=0.430 ηp^2=0.042	-6.67 to 3.00
Transverse angular velocity	11.60 (0.92)	10.63 (0.98)	-8.4	p=0.350 ηp^2=0.058	-3.12 to 1.17	57.51 (5.10)	50.36 (5.63)	-12.4	p=0.226 ηp^2=0.096	-19.24 to 4.94
Coronal angular velocity	6.19 (0.35)	5.94 (0.23)	-4.1	p=0.410 ηp^2=0.046	-0.89 to 0.38	23.78 (3.22)	21.19 (2.08)	-10.9	p=0.420 ηp^2=0.044	-9.26 to 4.07

Firm: Firm surface; Foam: Foam surface; % Diff: Percentage difference; 95% CI: 95% confidence interval; Dom: dominant; *: significant difference

RESULTS

The average age of the patients was 29 ± 9.3 years, height= 175.3 ± 9.4 cm, mass= 78.5 ± 18.9 kg, and Body Mass Index= 25.4 ± 5.6 kg/m^2. Of the 17 participants, three were left-limb dominant. Out of the seventeen participants, one failed to perform the SLS-foam after two failures on the foam surface resulting in an incomplete data set for this individual, no participants reported any notable fatigue. No interactions were seen between leg dominance and surfaces. Significant differences were seen between the firm and the foam surface for all the angular velocity measurements (p<.050), with the exception of the average angular velocity of the shank in the coronal plane (p<.050).

Comparing the surfaces, both the greatest effect size and percentage difference were found in the pelvis angular velocity data. With the average pelvis transverse plane angular velocity showing an effect size of ηp^2=0.826 with a corresponding percentage difference in maximum coronal plane angular velocity of 364% (Table 1).

The average calcaneus angular velocity in the coronal plane had the third highest effect size (ηp^2=0.817) and the second highest percentage change (298%) (Table 1). Whereas the shank showed the greatest effect size in the maximum M-L acceleration (ηp^2=0.779) which also showed the greatest linear acceleration, with a percentage change of 88% (Table 1).

With the exception of the anteroposterior linear acceleration of the shank, both effect size and percentage change showed that all average linear acceleration values were less sensitive to detect change between surfaces when compared to the average angular velocity values of the same body segments (Table 1). The same findings were seen for the maximum medio-lateral linear acceleration of the shank values for the percentage change, but not for the effect size (Table 1).

The greater effect size for all the maximum linear acceleration measures reflected their greater ability to detect difference between surfaces when compared to the average values of the corresponding axes and body segments (Table 1). In addition, the percentage change showed the same results for both acceleration and angular velocity with higher percentage change for the maximum values when compared to the average values (Table 1).

The comparison between the dominant and non-dominant leg, showed significantly greater values in the dominant versus the non-dominant sides (p<.050). These included the average and maximum proximal-distal accelerations of the calcaneus, the shank, and the mediolateral pelvis acceleration (Table 2). Within both average and maximum values, the three greatest effect size were seen for the calcaneus (ηp^2=0.589; ηp^2=0.381) and shank proximal-distal acceleration (ηp^2=0.781; ηp^2=0.405) and pelvis medio-lateral acceleration (ηp^2=0.371; ηp^2=0.387).

However, the side to side comparison also showed that the greatest percentage differences were seen in the average coronal plane angular velocity of the shank (-91.52%), the average mediolateral acceleration (-50.00%) and for the maximum mediolateral acceleration of the pelvis (-43.05%) (Table 2).

DISCUSSION

This work aimed to explore which were the most sensitive measures from IMUs when assessing postural stability and side to side differences. This highlighted that angular velocity measures appear to be more sensitive to different levels of challenge during static balance tasks when compared with acceleration measures, and that maximum values could be more sensitive compared to average values. When comparing different body locations and planes, neither a location nor a plane could be put forward as being more sensitive than another for the assessment of postural stability. Our results also showed that the dominant leg has greater acceleration and angular velocity values when maintaining balance compared to the non-dominant leg, which highlighted differences within the pattern used between leg to maintain postural stability and this mostly within the linear acceleration measures.

Previous studies have found that RMS of the resultant, anteroposterior and mediolateral linear acceleration are sensitive to distinguish different challenging static balance tasks whilst the same measurement in a vertical direction were not sensitive (8,9,13,14). These studies were able to determine differences between tasks, however the sensitivity of these measures in terms of effect size and percentage change between different tasks and surfaces had not been fully explored. In addition, no specific body location investigated had a greater effect size or percentage change. This could indicate that no one predominant strategy for maintaining postural stability from any particular body segment was present. Moreover, while all the other studies took the average RMS values into account for distinguishing different challenging static balance tasks (9,13), our studied identified that maximum RMS values may be more suitable for measuring differences. We hypothesized that angular velocity measures may provide a better reflection on the differences between different balance challenging tasks than linear accelerometer measures, due to the concept that segment angular velocity directly reflects the joint control and stability through the action of muscle control on rotational movement and not translation. This is not a new concept, Richards et al (2003) determined that angular velocity was more sensitive than joint angles and timing when determining the differences between paretic and non-paretic sides in Stoke survivors during different lower limb tasks, however it is now possible to perform such assessments in real time using data from IMUs (15).

It has also been shown that the assessment of segmental excursion of the lower limb might decrease the risk of reinjury by allowing better decision making regarding the return to sport with a particular emphasis on side to side differences (1). While Kim et al. compared linear acceleration measurements form IMU's positioned on the legs during SLS, the same comparison using angular velocity measurements has, to the best of the authors' knowledge, never been explored.

The results of our study show, relative to the side to side differences, that with both the average and maximum RMS values, the three greatest effect sizes were seen for the

calcaneus and shank proximal-distal acceleration and pelvis medio-lateral acceleration. In addition, the greatest percentage differences, were seen in the average coronal plane angular velocity of the shank, and in the average mediolateral acceleration and maximum mediolateral acceleration of the pelvis.

In comparison to the results of Huurnink et al. (2014) who did not find significant difference between side to side body segment angular velocity measurements while performing the SLS we found one significant difference in the pelvis coronal angular velocity maximum RMS values. This difference between the results might be explained by the differences in the body segments analyzed, and in the system used to acquire data. This may highlight that IMUs system and pelvis measures are more sensitive to detect between leg differences than optoelectronic camera system and ankle and hip measures. In addition, significant differences in the pelvis coronal angular velocity maximum RMS values, have recently been shown by Budini et al. (16), who used IMU angular velocity measurements and highlighted their potential value in the assessment of rehabilitation with regards to differences between lower limb during the Y balance dynamic stability test.

There were several limitations of this study. There were large differences in the two balance challenges between the conditions, whether the measures identified would be able to detect more subtle changes in surface is currently unknown. This work only considered healthy subjects and the sensitivity of these measures to detect clinically important differences when compared to injured athletes is also unknown. The results of this study only concerned a specific task and its utility with other balance tests needs further exploration. Finally, performing only one repetition of the single-leg stance for each foot may could not be representative of multiple repetitions of this task. Finally, the small sample size might have impact on the precision of the results and the sampling variability.

CONCLUSIONS

Overall segment angular velocity measures appeared to be more sensitive to detect differences between different balance-challenging tasks when compared to the linear acceleration measures. However, when considering the between leg differences linear acceleration measurement seems to be more sensitive to detect change than angular velocity measurements. Therefore, IMUs measures have the potential to be used within clinical assessment, and segment angular velocity measures might be more appropriate in the evaluation of postural stability and that linear acceleration measures, although linear acceleration are still useful in the assessment of between leg differences.

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