Research article

A novel approach to measuring wobble board performance in individuals with chronic ankle instability

Andrea Fusco [a,b], Philip X. Fuchs [a,b], Marianna De Maio [a], Herbert Wagner [b], Cristina Cortis [a,*]

[a] Department of Human Sciences, Society and Health, University of Cassino and Lazio Meridionale, Via S. Angelo - Località Folcara, 03043 Cassino (FR), Italy
[b] Department of Sports Science and Kinesiology, University of Salzburg, Schlossallee 49, A-5400 Hallein (Rif), Salzburg, Austria

ABSTRACT

Computerized wobble boards (WB) are used to objectively assess balance in healthy and chronic ankle instability individuals. As in field setting health professionals might not own WB, objective evaluations are not always feasible. Therefore, the aim of this tudy was to investigate the contribution of sagittal plane joints angular-displacement and anthropometrics to predict equations to estimate WB performance by portable two-dimensional motion analysis (2D-MA) and cross-validate the developed equations in chronic ankle instability individuals. Thirty-nine healthy and twenty chronic ankle instability individuals stood on a WB in single stance position. The balance test consisted of three 30s trials per limb keeping the platform flat at $0°$. Trials were video recorded, and three time-segments joints angular-displacement analyzed with 2D-MA: segment 1 (T1) including 30s data, segment 2 (T2) from second 0 to 10, segment 3 (T3) only the first 5s. Mixed regression for multilevel models was used to estimate WB performance for each time-segment and to examine limb differences for the predicted WB performance in chronic ankle instability sample. The accuracy of the equations to detect injured limbs was calculated via area under the curve for receiver operating characteristic. Ankle and knee angular-displacement parameters, body height and lower limb length were the major predictors of WB performance for the extrapolated models ($p < 0.05$; $R^2 = 0.83–0.56$). The measured WB performance and T1 model showed significant ($p < 0.05$) performance differences between the injured and uninjured limbs. Receiver operating characteristic analysis showed an asymptotic significance of 0.03 for T1 equation with area under the curve of 0.70. The proposed models provide different methods to quantify the performance and accurately detect the injured limb in individuals with unilateral chronic ankle instability, when measuring balance via WB might not be feasible. App-makers may use the equations to provide an automatic all-in-one system to monitor the performance status and progress.

1. Introduction

The ability to integrate sensory inputs from several receptors to determine human's movements and position in space (i.e., proprioception) plays a key role in balance control [1]. Dynamic and continuous information from the vestibular, visual and proprioceptive systems are required to provide neuromuscular adjustments essential to keep the human body center of mass within the base of support. Balance control is directly influenced by the sensorial information received and indirectly by previous injuries [2], range of motion (ROM) [3], anthropometric characteristics [4], side-general and site-specific limb effects [5], and training [6, 7]. Among several documented injuries that might influence the balance performances, ankle sprains and consequent residual symptoms such as the development of chronic ankle instability (CAI), are the most recurrent in sports, military and occupational settings, and generally in physically active people [8, 9].

Balance control is often assessed to evaluate changes after rehabilitation training intervention, deficits from previous ankle sprains and detect risk of reinjury in individuals with CAI [7, 10, 11]. Among the different methodologies, wobble boards (WBs), unstable platform generally used for proprioceptive training and rehabilitation protocols [6, 12, 13], have been recently computerized with accelerometers and connected to a computer to show reliable real-time data on balance in healthy and CAI individuals [2, 14, 15]. These systems proved to be easy

to set up, collect and interpret data and offer the potential to monitor individuals' dynamic balance during large-scale evaluation, also in field settings. Moreover, three-dimensional (3D) motion analysis has been used to objectively quantify the WB performance and its progression [16, 17]. However, due to costs and expertise required for data collection and analysis, 3D motion analysis might not be feasible in more practical settings, and therefore, two-dimensional (2D) motion analyses [18] are preferred.

Despite the lack of precision and ability to capture rotations, 2D motion analysis could provide a practical method of evaluating sagittal plane joint displacement for assessing gross movement shift during laboratory and field testing, and therein risk of lower extremity injury [19]. Therefore, 2D motion analysis video systems might be used safely in clinical practice as they are portable, time and cost effective, and require little rater training [18].

Although specificity, affordability, and transportability are key factors for making the data collection accurate and precise, in some cases physical therapists, athletic trainers, practitioners and health scientists might not own a computerized WB, or simply, the device might have some technical problems. Consequently, computerized WB could not always be the most adequate tool to measuring dynamic balance, thus making the 2D motion analysis a potential solution to overcome these limitations. It could be hypothesized that 2D motion analysis system might be an accurate and precise method to estimate the computerized WB performance in healthy and CAI individuals, thus providing an alternative tool for athletic trainers and physical therapists to evaluate the dynamic balance in field setting. Therefore, the aims of this study were: (a) to investigate the contribution of sagittal plane joints (hip, knee and ankle) angular-displacement and selected anthropometrics on a computerized WB performance, (b) to predict useful equations to estimate the WB performance by using 2D motion analysis system, and (c) to cross-validate the developed WB equations in individuals with unilateral CAI.

2. Methods

2.1. Experimental approach to the problem

Computerized WBs have been recently considered useful, precise, and reliable device for balance assessment showing intraclass correlation coefficients (ICCs) ranging from 0.65 to 0.89 in healthy subjects [14] and 0.58 and 0.84 in CAI individuals [2]. However, to fulfill the lack of accurate surrogate methods that might substitute computerized WB during balance evaluation in clinical practice, in this study a novel approach was favored. Therefore, the concurrent use of 2D motion analysis was chosen to develop equations for estimating the WB performance because highly affordable and reliable. According to previous studies [2, 15], the one leg stance was adopted for the evaluation of the WB performance because it is a common and challenging method widely used to discriminate between healthy and CAI subjects. During one experimental session a total number of six WB tests trials were performed after a familiarization period. To avoid potential fatigue, subjects were required to refrain from any moderate-to-vigorous physical activity for at least 24 h before the experimental session. All data were collected during morning sessions because diurnal patterns have been observed in dynamic balance performances [20]. Furthermore, to avoid potential effects on performances due to dehydration, participants drank water ad libitum during before and during the experimental sessions [21].

2.2. Participants

Thirty-nine healthy and twenty unilateral CAI recreationally active (engaging in at least 3 days a week of moderate-to-intense physical activity) young adults provided written informed consent to participate in the study carried out in accordance with the Declaration of Helsinki for Human Research of 1964 (last modified in 2000). The study was approved by the local Institutional Review Board (approval number: 14357.2019.06.18). The healthy participants were voluntary recruited from the local community and selected to sufficiently cover a wide range of anthropometric characteristics. They were included if self-reported: no previous injuries, fracture, or surgery of either ankle; no cerebral concussions, lower extremity injuries, vestibular and visual disorders for 3 months before testing; no ear infection, upper respiratory tract infection at the time of the study; no prior balance training. Unilateral CAI participants were selected [22] if they self-reported: at least one unilateral ankle sprain, but none within the past 6 weeks; multiple (more than 3) episodes of unilateral ankle giving way within the past 12 months; no previous fracture or surgery of either ankle; no cerebral concussions, lower extremity injuries, visual and vestibular disorders for 3 months before testing; no ear infection, upper respiratory tract infection at the time of the study; no prior balance training.

Body mass and height were measured by means of a scale with integrated stadiometer with a precision of 0.1kg and 0.1cm (Seca, model 709, Hamburg, Germany), and body mass index (BMI) calculated. Lower limb length was measured from the anterior superior iliac spine to the most distal part of the medial malleolus by using a tape measure while the subject laid in supine position. Limb dominance was also determined by asking the favorite foot to kick a ball.

2.3. Procedures

The WB performance was assessed via a computerized proprioceptive board (Balance Board WSP, Rome, Italy; 40cm diameter with a half plastic sphere of 6cm height and 20cm width; maximal tilt angle = 20°) equipped with a triaxial accelerometer (Phidget Spatial 0/0/3 Basic 1041, Calgary, Canada). After a 3-minute familiarization, followed by 1-minute rest in sitting position, the participants stood barefoot on the WB in a single leg stance position, finding a comfortable and central position with the knee slightly bent and keeping the hands on the hips. The balance test consisted of three 30-second trials per limb with 1-minute sitting rest in between. Starting limb was randomly chosen. During the test each subject was asked to focus on the motion marker (diameter = 6mm) displayed on the monitor (1920 × 1080 resolution screen) placed at eye level 2-meter in front of them and to keep it inside the target zone (diameter = 6.5cm) as long as possible. The target zone was represented by a circle showing the 0° tilt angle measured by the triaxial accelerometer. The boundaries of the motion marker and target zone were standard for all participants during each trial. The data collected for analysis was the time (s) spent by the motion marker inside the target zone, which expresses the time the subject spent on the platform keeping it flat at 0°. Visual markers were applied by the same expert researcher to the participants' base of the fifth metatarsal, lateral malleolus, lateral joint line of the knee, anterior superior iliac spine and acromion process. All trials were recorded by a video camera (Sony Camcorder HDRCX290/B; Sony, Minato, Tokyo, Japan) laterally fixed at 2.30 m from the participants and 1 m above the ground. One researcher recorded the test trials, imported the videos on a motion analysis software (Dartfish Team Pro 5.5; Dartfish, Fribourg, Switzerland) and calculated hip, knee and ankle angular-displacement data in the sagittal plane (Figure 1).

On each video, the same trained researcher measured joint angles from the beginning to the end of the test trial using the visual markers as references. Hip angular-displacement (Figure 1a) was measured as the angle between the acromion process and lateral joint line of the knee with the greater trochanter serving as the fulcrum. Knee angular-displacement (Figure 1b) was measured as the angle between the greater trochanter and lateral malleolus with lateral knee joint serving as the fulcrum. Ankle angular-displacement (Figure 1c) was measured as the angle between a line from the lateral knee joint and the base of the fifth metatarsal with the lateral malleolus being the fulcrum. The recorded videos of the subject's performance were analyzed at 25 frames per second. To reduce the amount of time for video analysis by athletic trainers, physical therapists and health professionals during field testing,
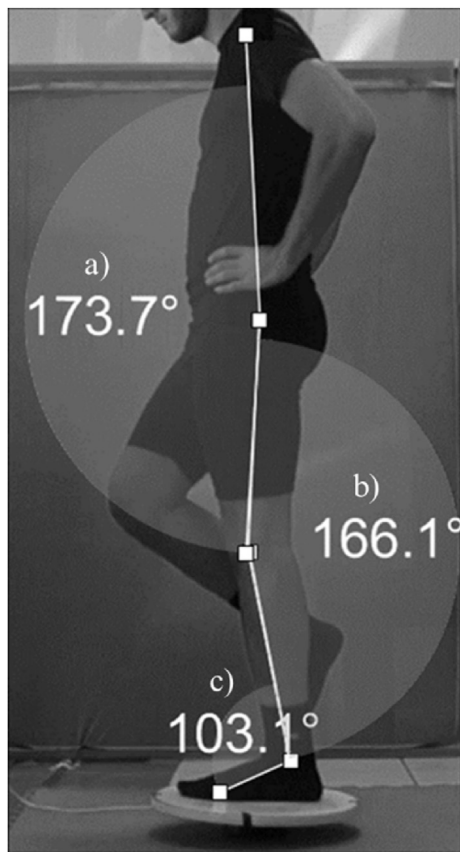
Figure 1. Example of hip (a), knee (b) and ankle (c) joint angles data in the sagittal plane.

three different time segments of the WB tests were analyzed for further statistical analysis. Segment one (T1) included 2D motion analysis data of all 30 s of the WB trial (from second 0 to second 30). Segment two (T2) included motion analysis data from second 0 to second 10. For segment three (T3) only the first 5 s of the WB trial (from second 0 to second 5) were video analyzed.

2.4. Statistical analysis

Data were analyzed using STATA 14 (StataCorp LP, Texas, USA). Normal distribution was verified by the Shapiro-Wilk test. Means, variance, standard deviations (SD) and range were calculated for all variables. For the WB performance, mean, variance, SD and range were calculated using all video analyzed frames as single data point for each trial and subject. Multilevel mixed regression models were created to predict equations to estimate the WB performance for each time segment video analyzed. The healthy participants were used as random effects with repeated measurements of WB performance for each subject. Bryk/Raudenbush R-squared (R^2) values and root mean squared error (RMSE) for each model were calculated. ICCs for multilevel models were also estimated. The association between measured and predicted WB performance, evaluated by calculating the Pearson's correlation coefficients (r) for each mode, was used as a measure of precision, whereas the bias-correction factor (C_b) was used as measure of accuracy. Subsequently, the Lin's concordance correlation coefficient (rho_c) was calculated as the product of r and C_b. Bland-Altman plots showing level of agreement and regression line fitting the paired differences to the pair-wise means were plotted to assess non-constant bias.

To cross validate the developed equations a subsample of twenty unilateral CAI individuals performed a single CWB trial for each leg (injured and uninjured). Multilevel model regression was performed to

examine potential differences between injured and uninjured limbs for the measured and predicted WB performance in the unilateral CAI sample. Participants were considered as random effect, whereas the limbs were treated as fixed effect. The models were fitted using the residual maximum likelihood to account for the small sample. The contrast method was used to test whether the measured and predicted WB performance were identical between limb and extrapolated equations. The contrast method tests include ANOVA-style tests of the main effects used to make comparisons against the reference (measured WB performance and uninjured limb). To provide meaningful analysis for comparisons from small groups, Cohen's effect sizes (ES) were also determined. An ES less than 0.2 was considered trivial, from 0.2 to 0.5 small, greater than 0.5 to 0.8 moderate, and greater than 0.8 large. Bonferroni post-hoc tests were used for multiple-comparison adjustments across all terms. Lastly, the accuracy of the predicted WB measures in detecting injured limbs in the CAI individuals was calculated using the area under the curve (AUC) for receiver operating characteristic (ROC) curve. An academic point scale was used to classify the accuracy of the AUC for discriminating between injured and uninjured limb: fail (0.00–0.59), poor (0.60–0.69), fair (0.70–0.79), good (0.80–0.89), and excellent (0.90–1.00). The significance level was set at $P < .05$.

3. Results

Participants descriptive characteristics and joint angle average values are presented in Table 1.

Three multilevel regression models were created using the WB trial performance of the healthy participants as dependent variable (Table 2).

In the first model (T1), lower limb length and ankle angle parameters (mean, variance, SD and range) were used as independent variables, with significant ($p < 0.05$) effects for all variables (ICC of 0.22). Analyzing model T1 (Table 2A), the following equation to estimate the WB performance was extrapolated:

T1 = 36.56276 + 0.127184*ankle mean (°) + 0.4046644*ankle variance (°) - 4.529743*ankle SD (°) - 0.2324548*ankle range (°) - 0.2372182*lower limb length (cm).

In the second model (T2), body height, ankle angle parameters (mean, variance and SD) and knee angle SD had significant ($p < 0.05$) effects (ICC of 0.26). Accordingly, the following equation was extrapolated from the model T2 (Table 2B):

T2 = 36.7864 + 0.1738654*ankle mean (°) + 0.4629237*ankle variance (°) -5.220193*ankle SD (°) + 0.5952131*knee SD (°) - 0.1622368*body height (cm).

Finally, only the first 5 s of the WB trial (from second 0 to second 5) were video analyzed for developing the third model (T3). Lower limb length and ankle angle parameters (mean, variance and range) were significant ($p < 0.05$), with an ICC of 0.36. Therefore, the following equation to estimate the WB performance was extrapolated (Table 2C):

T3 = 31.8308 + 0.1619749*ankle mean (°) + 0.1978885*ankle variance (°) - 0.6410204*ankle range (°) - 0.3059346*lower limb length (cm).

Bland-Altman plots and fitted regression lines with rho_c coefficients for the healthy and unilateral CAI individuals are shown in Figure 2.

The mixed effects linear regression analysis showed significant differences between injured and uninjured limb and between the measured and predicted WB performance ($F_{7,133} = 8.80$, $P < .0001$; Figure 3).

Comparisons after Bonferroni corrections showed significant differences between the injured and uninjured limb for the measured WB performance ($P < .001$, ES = 1.10) and the T1 model ($P = .012$, ES = 0.65). Furthermore, significant differences were found for the injured limb between the measured WB performance versus T2 ($P = .003$, ES = 0.64) and versus T3 ($P < .001$, ES = 0.77). The predicted models and measured WB performance did not show significant differences between the uninjured limb of the CAI sample.

The ROC curve analysis showed an asymptotic significance of 0.03 only for the T1 extrapolated equation with an AUC of 0.70 (Figure 4). The

Table 1. Descriptive characteristics and joint angle average values of the healthy and unilateral chronic ankle instability (CAI) individuals.

Descriptive characteristics	Healthy individuals			CAI individuals		
	Mean	SD	Range	Mean	SD	Range
Age (years)	23.1	2.4	10	23.5	1.5	6
Mass (kg)	64.6	10.4	34	67.3	12.9	49
Height (cm)	167.3	8.1	39	167.8	9.9	32
Lower limb length (cm)	78.8	5.3	25	78.0	6.3	23
Body mass index (kg·m^{-2})	22.9	2.8	10.2	23.9	4.1	15.4
	All individuals' healthy limbs			**CAI injured limbs**		
Joint angle average values*	**Mean**	**SD**	**Range**	**Mean**	**SD**	**Range**
Ankle (°)	104.1	2.1	12.5	104.6	2.8	15.4
Knee (°)	163.2	1.9	9.5	164.6	2.4	13.8
Hip (°)	170.0	2.2	11.1	170.3	3.3	16.3

* Data represent the average of all video analyzed trials using frames as single data point for each subject.

best cutoff values identified were 19.5s (sensitivity = 0.55; 1-specificity = 0.20) 20.5s (sensitivity = 0.60; 1-specificity = 0.35) and 21.5s (sensitivity = 0.75; 1-specificity = 0.45).

4. Discussion

The aims of our study were to investigate the contribution of sagittal plane joints (hip, knee and ankle) angular-displacement and selected anthropometrics on a computerized WB performance, to predict useful equations to estimate the WB performance by using a 2D motion analysis system, and to cross-validate the developed WB equations in individuals with unilateral CAI. Our main findings were that the ankle and knee angular-displacement parameters, body height and lower limb length were the major predictors of the WB performance. Furthermore, the extrapolated models accurately predicted the WB performance in healthy individuals, whereas only the T1 model was able to accurately detect WB performance differences between the injured and uninjured limb in individuals with unilateral CAI.

The first relevant finding from our study showed that the ankle, independently from the time-segment video analyzed, and knee angular-displacement played major roles on the WB performance and the accuracy of the predicting models extrapolated. This result is in line with previous studies [16, 23], which have shown that the control of standing balance during single limb tasks relies on the control of the ankle with increasing contributions of proximal joints as the balance demands become more challenging. Regarding the selected anthropometrics, only body height and lower limb length had an influence on the WB performance. Previous studies reported that body height and lower limb length should be considered during balance assessment [4, 24], while mainly focused on reaching tests for normalization purposes. To the best of our

Table 2. Mixed regression models between wobble board test performance and independent variables.

	Coef.	SE	z	P>|z|	[95% CI]	
A) Wobble board test (T1)						
Ankle mean (°)	0.127184	0.0515616	2.47	.014	0.0261252	0.2282429
Ankle variance (°)	0.4046644	0.1095888	3.69	<.001	0.1898742	0.6194546
Ankle SD (°)	-4.529743	0.8805929	-5.14	<.001	-6.255673	-2.803812
Ankle range (°)	-0.2324548	0.0925858	-2.51	.012	-0.4139197	-0.05099
Lower limb length (cm)	-0.2372182	0.0661291	-3.59	<.001	-0.3668289	-0.1076076
_cons	36.56276	8.310829	4.40	<.001	20.27384	52.85169
R²: 0.83	**RMSE: 3.25**				**P: < .0001**	
B) Wobble board test (T2)						
Ankle mean (°)	0.1738654	0.0593022	2.93	.003	0.0576353	0.2900955
Ankle variance (°)	0.4629237	0.1317005	3.51	<.001	0.2047955	0.7210518
Ankle SD (°)	-5.220193	0.8799794	-5.93	<.001	-6.944921	-3.495465
Knee SD (°)	0.5952131	0.2694632	2.21	.027	0.0670749	1.123351
Body height (cm)	-0.1622368	0.0516359	-3.14	.002	-0.2634413	-0.0610323
_cons	36.7864	11.83021	3.11	.002	13.59962	59.97319
R²: 0.75	**RMSE: 3.62**				**P: < .0001**	
C) Wobble board test (T3)						
Ankle mean (°)	0.1619749	0.0646645	2.50	.012	0.0352349	0.288715
Ankle variance (°)	0.1978885	0.099032	2.00	.046	0.0037894	0.3919877
Ankle range (°)	-0.6410204	0.1708659	-3.75	<.001	-0.9759114	-0.3061294
Lower limb length (cm)	-0.3059346	0.0961308	-3.18	.001	-0.4943475	-0.1175217
_cons	31.8308	10.92265	2.91	.004	10.4228	53.23881
R²: 0.56	**RMSE: 3.74**				**P: < .0001**	

T1 = Time-segment one; T2 = Time-segment two; T3 = Time segment three; SD = Standard deviation; _cons = Intercept; coef. = Coefficient; SE = Standard errors; CI = Confidence Interval; R2 = Bryk/Raudenbush R-squared; RMSE = Root mean squared error.
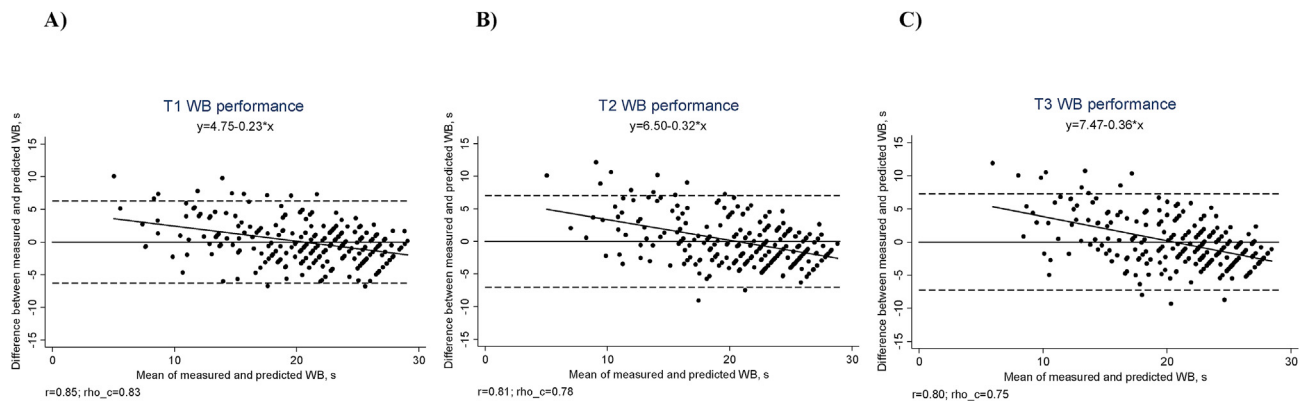
A)

B)

C)

Figure 2. Bland-Altman Plots for A) T1, B) T2, and C) T3 extrapolated wobble board performance models. Difference between predicted and measured wobble board performance is plotted against the mean of the respective measurements. Horizontal black line indicates the average of the differences, whereas the dashed lines show the upper and lower 95% limits of agreement. Black fitted linear regression line is also displayed. T1 = Time-segment one; T2 = Time-segment two; T3 = Time segment three; WB = Wobble board; r = Pearson's correlation coefficient; rho_c = Lin's concordance correlation coefficient.

knowledge, there are no studies that investigated the direct impact of lower limb length on balance performances assessed on computerized WB. However, our results regarding body height are in line with Greve et al. [25], which demonstrated that body height had moderate correlation with balance performances evaluated on a Biodex balance system. Therefore, our findings confirm what has been previously reported by Berger et al. [26] and Alonso et al. [27], which stated that ankle displacements increased with body height. This is further explained by the "inverted pendulum" theory [28]. According to the theory, during upright position the human body can be compared to an inverted pendulum system rotating around the ankle joint, thus the anthropometrics, especially body height, could be affected by the total load of movements occurring at the top of the inverted pendulum [29]. Therefore, an increase of body height, could lead to an increased ankle torque essential to keep postural balance particularly during single leg stance on unstable platform [25, 26, 27].

Three different time-segments were video analyzed by the same trained researcher in order to develop useful equations for predicting the WB performance. The R^2 statistics, the % variation in measured WB explained by the model, ranged from 56% (T3) to 83% (T1), with an absolute difference between the predicted and measured WB performance ranging from 3.25-3.74 s. Alongside the video analysis of the overall WB performance (T1), for practical reasons we choose, a priori, to develop equations for the first 10 s (T2) and 5 s (T3). Clinically, the ability to quickly reproduce reliable surrogate measurements is crucial, especially during large scale evaluations. Therefore, to estimate the

adequacy of the models, the ICCs for multilevel models were calculated. Numerical value of this index, ranging from 0.22 to 0.36, indicate that multilevel modeling is a suitable model to analyze the existing data and multilevel analysis can better present results compared to simple regression [30]. For example, the T2 ICC of 0.26 would suggest that 26% of the outcome variability depends on differences among individuals, whereas the remaining 74% depends on differences between the measurements made in the same subject.

Despite the measures of precision and accuracy were strong, the developed equations were slightly biased as we used a mixed model (with participants as random effect) method [31, 32]. In fact, visual examination of the Bland-Altman plots suggests that the differences between the measured and predicted WB performance were not constant, with the predicted models increasingly overestimating the measured WB performance [33]. By using the difference in WB
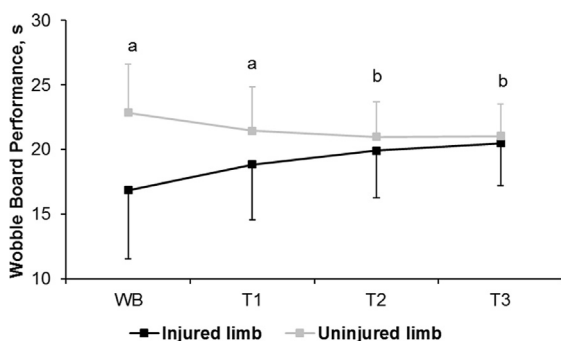

Figure 3. Means and standard deviations of meaured (WB) and predicted (T1 = 30s; T2 = 10s; T3 = 5s) wobble board performance across injured and uninjured limbs in the chronic ankle instability sample. a: significantly ($P < .05$) different from the uninjured limb; b: significantly ($P < .05$) different from the measured (WB) wobble board performance.
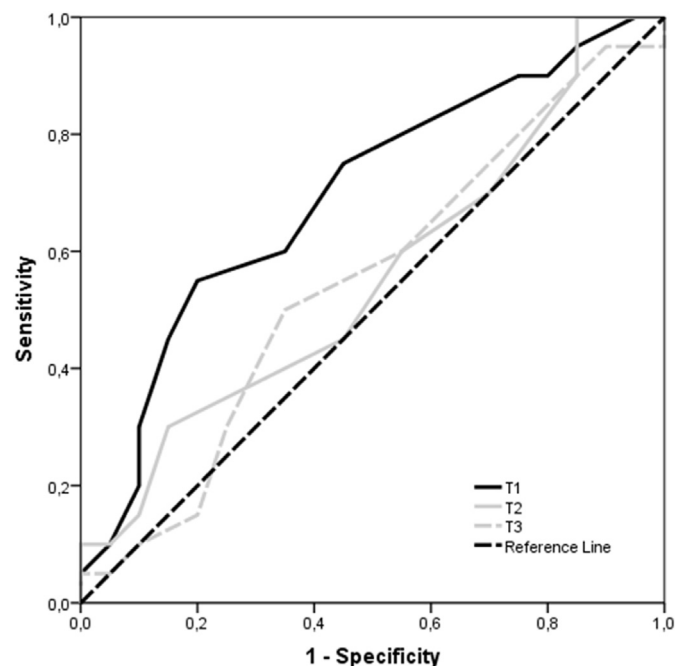

Figure 4. Receiver operating characteristic (ROC) curve for the extrapolated wobble board performance models (T1 = solid black; T2 = solid grey; T3 = dotted grey) indicating sensitivity and 1– specificity tradeoff are shown relative to the reference line (dotted black), which indicates that a test performed no better than random.

performance as a dependent outcome variable and the mean between the two methods of measurements as an independent predictor in a linear regression for each model, regression line with slope ranging from -0.23 to -0.36 were obtained. For example, the predicted T2 equation, on average, overestimated the measured WB performance by 0.32 s for each second increase in the measured WB performance. A biphasic trend was also evident in all Bland-Altman plots. Therefore, according to the regression line, the predicted values on the y-axis were positive at the lowest mean value of the two measurements on the x-axis, and the values on the y-axis were negative at the largest difference between the two methods. This suggests that the extrapolated models overestimate the measured WB performance when the magnitude of the measurement is large, but on the other hand, underestimate when the magnitude is small [34].

The consistency and precision of the extrapolated models as indirect methods for WB assessment in healthy limbs is clearly supported by the results. In fact, no significant differences were demonstrated between the measured WB performance and extrapolated models for the uninjured limb of the unilateral CAI sample. On the other hand, the T2 and T3 models were unable to successfully and accurately detect limb differences in individuals with unilateral CAI. Interestingly, the cross-validation of the developed equations in individuals with unilateral CAI showed that the T1 extrapolated model was able to successfully and accurately detect limb differences in individuals with unilateral CAI, alongside the measured computerized WB outcome, which have shown to be in line with previous studies [2, 15]. The accuracy of the T1 extrapolated equation is further strengthen by the significant AUC of 0.70, which is considered to be fair. The best cutoff values ranged from 19.5s to 21.5s, which are similar to the one reported in a previous study of 18.5s [2]. Therefore, based on the cutoff value of 20.5s, the 60% of the injured limb (true positive) would be correctly classified as injured, whereas the 35% of the uninjured limb would be incorrectly identified as injured (false positive).

Despite the meaningful results of our investigation, some limits need to be acknowledged. Our sample was limited to healthy young adults and participants with unilateral CAI, and therefore other populations, such as older adults or other clinical populations, could have different predictors and results for the WB performance. Secondly, as feedback can enhance neuromuscular control, it is possible that visual feedback provided when showing real time performance, could have affected the influence of anthropometrics and joints angular-displacement parameters on the WB performance [35]. Therefore, it should be determined whether the predicted models would have similar precision and accuracy with or without visual feedback. As this study analyzed only the sagittal plane, future researches should investigate other planes of motion as well as other joints. Lastly, the analysis of human movement using 3D or 2D motion analysis system is prone to instrument and observer errors, such as the identification of anatomical landmarks. Therefore, future studies should investigate the interrater reliability and consistency of such approach, as well as the cross-validation of the predicted models with other clinical populations.

5. Conclusions

Ankle and knee angular-displacement parameters, body height and lower limb length were the major predictors of the WB performance and played major roles on the accuracy of the extrapolated models. The equations may provide different methods to quantify the WB performance and accurately detect the injured limb in individuals with unilateral CAI, when measuring balance via computerized WB might not be feasible. Therefore, this could help physical therapists, athletic trainers, practitioners and health scientist to quickly assess the WB performance. Furthermore, app makers may use the equations to provide an automatic all-in-one system to monitor and document the WB performance status and progress.

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