Can the Nintendo WII Balance Board be Used for a Reliable Assessment of the Initiation of Gait?

Can the Nintendo WII Balance Board be Used for a Reliable Assessment of the Initiation of Gait?

Federica Verdini, *Member, IEEE*, Andrea Tigrini, *Student Member, IEEE*,
Sandro Fioretti, *Member, IEEE*, and Alessandro Mengarelli, *Member, IEEE*

***Abstract*—The Nintendo WII balance board (NWBB) is appreciated not only as a gaming device but also as an alternative to laboratory grade force plate in clinical and human motion research applications. Despite its validity during postural and quasi-static motor tasks has been evaluated in several studies, no hints were provided about its usability during gait initiation for the anticipatory postural adjustments analysis. In this study the validity of the NWBB was assessed by comparing temporal and spatial parameters from center of pressure trajectories with those obtained from a dynamometric force plate. The similarity between the trajectories was confirmed by the low values of root mean square error. The percentage errors in spatial parameters resulted under 10% for the whole trajectory and under 15% for the anterior/posterior and medial/lateral component respectively. Bland-Altman plots showed errors equally distributed around the mean difference, without a significant proportional tendency. Consistency and agreement between measures, verified by the high values of intra-class correlation coefficients, were further confirmed by temporal parameters characterized by limited errors, lower than 14%, for each gait initiation phases. Findings of the present study confirm the usability of the NWBB not only for static but also dynamic tasks and can contribute to enhance the use of such device for investigating the initiation of gait in clinical and not-specialized contexts.**

***Index Terms*—Initiation of Gait, Anticipatory Postural Adjustments, Center of Pressure, Force Plate, Nintendo Wii Balance Board**

I. Introduction

In recent years, gaming systems, as Nintendo WII Balance Board (NWBB) or Kinect MOtion CAPture systems (KMOCAPs), encountered a growing interest in contexts different from the entertainment, such as rehabilitation and sport [1], [2]. The low cost and the easiness to use, joined with a high level of acceptability, make such kind of technology suitable for clinical and sport training applications [3], [4]. Though in gaming applications the macroscopic features of the subject's motion are required for interacting with the environment, in clinical or sport use, the measurement accuracy becomes a fundamental requirement [5]. While KMOCAPs provide a kinematic representation of the human body through the measure of the virtual skeleton joints motion, the NWBB gives a direct measure of vertical force exchanged between feet and ground, thus allowing to obtain the Center of Pressure (CoP) trajectory. The accuracy and the reliability of measures obtained from such kind of devices, with respect to the gold standard for motion capture and kinetics measurements, have been object of several studies [2], [6]–[9]. With respect to this latter point, although the NWBB shows some limitations as a poor signal to noise ratio and the lack of shear forces or moments, its use during static posture has been validated with simultaneous measurements from the NWBB and laboratory grade force plate (FP) [7], [8] and its usability in clinical contexts has been assessed during instrumented static postural test with healthy, elderly and subjects with disability [1], [10]–[13].

Although the role of the anterior-posterior (AP) and medial-lateral (ML) component of the ground reaction forces (GRF) appears less relevant during static posture task given the dominance of the vertical one, during non-static and non-vertically developed tasks, the lack of horizontal components in NWBB measures could limit its use [7]. This aspect entailed the need for a task-dedicated evaluation, as performed for squat, sit-to-stand and functional reach test [9], [14], [15]. The remarkable correlation between CoP trajectories and the limited fixed biases for related CoP parameters encourage to investigate the validity of NWBB measures for another clinically relevant test, as the initiation of gait (GI), during which the AP and ML components of CoP are relevant sources of information in anticipatory postural adjustments (APAs) characterization [16], [17].

During GI, the central nervous system (CNS) has to guarantee balance while the subject moves from the upright posture to a steady-state locomotion, passing from bi- to mono-podalic stance [16], [18]. Indeed, APAs in GI follow the intention of the movement and occur prior to the gross segmental movement by anticipating the changes of the stability boundary due to the single leg stance [16]. Before the movement onset, subject modifies its postural attitude by moving the Center of Mass as a result of the CoP shift, induced by the coordinated strategy at the ankle and hip joints. The action of the ankle dorsi-flexors and hip ab/adductors muscles determines the CoP movement during APA. As first, CoP shifts in lateral and posterior direction (APA1) toward the heel of the first rising foot (swing foot), and then displaces in lateral direction toward the stance foot (APA2) [16], [19]. The end of APAs is followed by a dynamic condition characterized by the forward CoP displacement, named locomotion phase (LOC), which lasts up to the toe-off of the stance foot [16], [18]. Given the high correlation between a poor control strategy during the transition from double- to single-leg stance and the risk of fall, the correct characterization of APA is a basic requirement for the early detection of possible deficits in balance maintenance [16], [20].

All authors are with the Department of Information Engineering, Università Politecnica delle Marche, 60131 Ancona, Italy.
Corresponding author: A. Mengarelli, email: a.mengarelli@pm.univpm.it

Although electromyographic systems and inertial sensors unit have been used for APAs characterization [17], [18], [21], the complexity of the instrumentation often requiring specialized staff, specific experimental procedures and structured environment, limits their use to well-equipped laboratory or clinical centers. In addition, they provide only an indirect measure of APAs-related parameters and thus the use of dynamometric force plates cannot be disregard since they allow to obtain the CoP trajectory from the direct measurement of the GRF. In this scenario, the NWBB is considered a valid cost-effective alternative to the FP in many experimental conditions related to CoP measurement and analysis [8]. Given the relevance of the GI in clinical context, this easy-to-use portable device could facilitate the analysis also in not high-specialized contexts [20]. Lee and co-workers [22] used NWBB during GI, by examining CoP parameters with the aim to find clinical signs in old adults at risk of fall with respect to healthy young people. Although they found high within-subjects reliability in ML component and a fair reliability in CoP total length, their findings are not supported by the analysis of NWBB validity with respect to the gold standard. Indeed to determine its suitability in the clinical practice, as pointed out by the same authors, is essential to evaluate if these differences are clinically relevant or due to sensor inaccuracy.

The present study is aimed to verify the validity of the NWBB as possible alternative to laboratory-grade force plate during GI by comparing spatial and temporal parameters computed from CoP time series, simultaneously recorded. Giving the clinical relevance of such motor task, the focus constitutes a novelty and the results can provide useful indication for the use of such device in research and clinical contexts.

II. Materials and Methods

A. Population

Ten healthy subjects (7 men and 3 women, their average ($\pm$ SD) age 22($\pm$1) years; body mass 60.0 ($\pm$1.5) kg; height 1.72 ($\pm$0.11) m) performed GI test. Any neurological and orthopedic disorders or other impairments that could potentially interfere with balance and gait were considered as exclusion conditions from the test. All the volunteers gave their written consent before their participation. The study was approved by the local ethics committee.

B. Task description

Each subject performed 10 repetitions of a GI trial for a total of 100 time-series. During each test the NWBB was placed over the laboratory grade FP and the subjects stood barefoot on them in a free and comfortable position for their feet and arms. A wood plate was positioned in front of the platforms to allow the subjects to perform the step and have a subsequent stance over a plane aligned with that of the platforms. Two auditory stimuli were provided to the subject: the first was given as a "warning signal (WS)" and the second, 2 s later, as a "go signal (GS)" (Fig. 1a). The subjects were instructed to take the step with a leg, named (*swing foot*, Fig. 1a) after hearing the GS. The foot landed on the wood plate while the contralateral foot, named (*stance foot*), moved after. The time instant corresponding to the toe-off of the (*stance foot*) was considered as the end of the trial (T_f in Fig. 1a). Subjects remained on the wood plate in upright posture for 10 s. Finally they went back to the initial position for the next step.

C. Data acquisition and analysis

Data were simultaneously acquired from NWBB (43.3$\times$23.8 cm, maximum vertical force range 1471 N) and from the laboratory grade FP (Bertec model 4060H, 40$\times$60 cm, maximum vertical force range 7000 N). Acquisition of both the platforms was triggered 10 s before the WS signal. FP data were acquired at 500 Hz. For NWBB, each strain gauge load sensor, located at the four corners of the board, measured the vertical components of GRF. Data were streamed through bluetooth connection (free software available on website https://wiimotephysics.codeplex.com/) and stored for the off-line processing (MATLAB, R2018b The Mathworks Inc). As already reported in [9], [23], the NWBB data showed the time jitter drawback, presenting an inconsistent sampling rate close to 100 Hz (average value of 92.2$\pm$8.7 Hz). To obtain a regular time-series the NWBB data were then interpolated and resampled at 500 Hz.

Each time-series were low-pass filtered at 20 Hz (zero-lag phase, 4th-order Butterworth filter). Based on the acquired force data, the anterior-posterior and medio-lateral coordinates of CoP, were obtained according with [23].

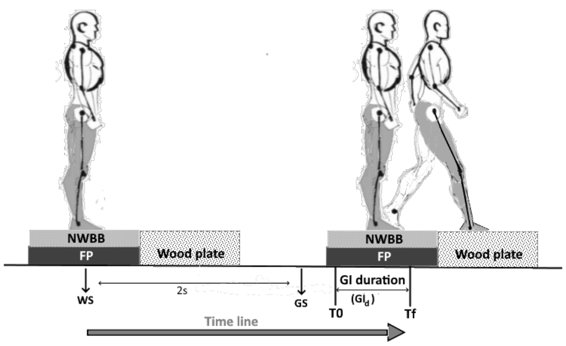
The average value calculated in the time-interval from WS to GS (2 s long, Fig. 1a) for the CoP time-series of both the devices, was subtracted to eliminated the bias. To automatically determine the onset APA detection [21], twice the standard deviation value from the AP component of CoP time-series was settled as threshold The time instant (T_0), at which the AP CoP component overcame this threshold value after GS event, was considered as the onset of GI trial while the end corresponded with T_f event (Fig. 1a).

Eventually, each trial of GI was analyzed by considering two phases, the postural and the locomotor (LOC) one [16], [18]. The postural phase, relevant to define APAs, was characterized as in [16] by:

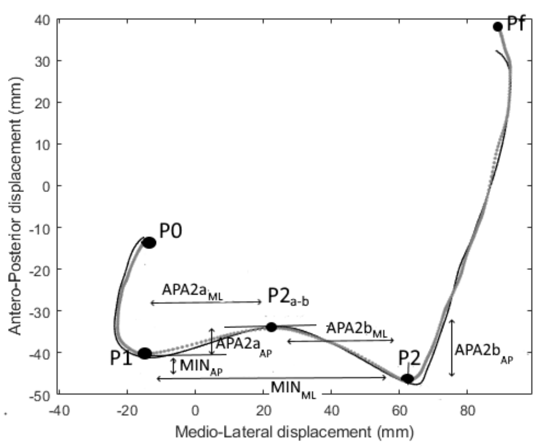
- APA1, between [P_0-P_1], (Fig. 1b), when the CoP moved backward and laterally toward the *swing foot*. This phase ended when the first minimum of the CoP in the AP direction was reached. Related length (APA1$_l$, in mm) was calculated together with its duration (APA1$_d$, in ms).
- APA2 started at the end of APA1 (P_1 at time instant T_1) and ended when the CoP reached the maximum medial displacement toward the *stance foot* (P_2 at time instant T_2). The absolute values of the distance between P_1 and P_2 in the AP and ML direction (MIN$_{AP}$, and MIN$_{ML}$, Fig. 1b) were calculated. This phase was further divided into two sub-phases based on the maximum of CoP displacement in AP direction (P_{2a-b}, Fig. 1b) between P_1 and P_2. The first sub-phase, APA2a, was between P_1 and P_{2a-b} while the second, APA2b, was between P_{2a-b} and P_2. Their lengths (APA2a$_l$, APA2b_l in mm, Fig. 1b), the absolute value of the distances in AP (APA2a$_{AP}$ and APA2b$_{AP}$ in mm, Fig. 1b) and in

ML direction (APA2a$_{ML}$, APA2b$_{ML}$ in mm, Fig. 1b) and their durations (APA2a$_d$, APA2b$_d$ in ms) were evaluated.

The LOC phase was defined between the P_2 and P_f (Fig. 1a) [16]. It was described through its length in mm (LOC$_l$) and its duration in ms (LOC$_d$). Eventually, the total duration of the GI (GI$_d$, Fig. 1a), evaluated between the time events T_0 and T_f, was calculated [16].



(a)



(b)

Fig. 1: (a) Representation of the assessment set-up with the NWBB,positioned over the FP and the wood plate, positioned in front of them. The swing leg, moving first, is grey colored. A time line shows the sequence of warning (WS) and the go (GS) signals together with the time-events, T0 and Tf, that identify the GI duration; (b) CoP trajectories obtained from FP (dashed line) and from NWBB (continuous line). P_0, P_1, P_{2a-b}, P_2 and P_f identify the GI events and APA2a$_{AP}$, APA2b$_{AP}$, APA2a$_{ML}$, APA2b$_{ML}$ labels specify distance parameters.

CoP similarity was evaluated in terms of root-mean-square error (RMSE) between FP and NWBB CoP trajectories during the entire GI and also for each considered sub-phase. The data from FP and NWBB were compared by computing absolute (AE) and percentage error (PE). The last one was calculated as the AE referred to FP measures, considered as the ground reference value. Linear regression analysis was applied to assess possible relationships between parameters obtained from both the devices. Agreement was analyzed through Bland-Altman plots and the reliability of measures between devices was evaluated through two-way mixed single-measure agreement and consistency intra-class correlation coefficients (ICC$_A$ and ICC$_C$), as in [7].

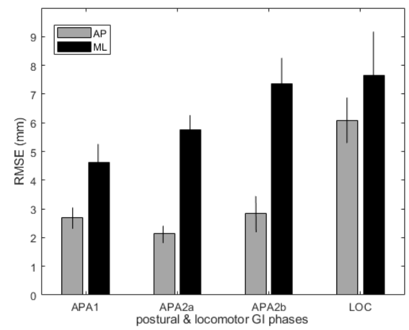


Fig. 2: Bar plot relative to the mean RMSE calculated for CoP displacement in the AP (grey color) and ML (back color) direction, during APA1, APA2a, APA2b and LOC phases.

III. Results

The CoP trajectories for NWBB and FP have showed a comparable trend in all the GI phases (Fig. 1b). Notably, the mean RMSE values, calculated on the whole GI duration, have resulted lower than 9 mm (8.3$\pm$5.2) for the AP displacement and lower than 13 mm (12.4$\pm$6.9) for the ML one, showing high similarity. This trend for the mean RMSE values has been also confirmed in the AP direction, resulting below 3 mm, and in the ML one, ranging from 4 to 8 mm, for each APA sub-phases (Fig. 2).

The validity of NWBB measures has been highlighted also by the limited PE (Table I) that resulted under 10% of FP values for length-related measures (APA1$_l$, APA2a$_l$, APA2b$_l$, and LOC$_l$) and under 16% for the distances-related measures (MIN$_{AP}$, MIN$_{ML}$, APA2a$_{AP}$, APA2a$_{ML}$, APA2b$_{AP}$ and APA2b$_{ML}$). All spatial measures have confirmed high level of agreement with ICC values higher than 0.95 with the exception for APA2b$_l$ phase, for which both the ICCs resulted equal to 0.88 (Table I).

Likewise, the linear regression analysis has highlighted a strong correlation between measurements, with r values not lower than 0.95 for all spatial parameters except for LOC$_l$ that has been equal to 0.88 (Fig. 3, 7, and 9). The absence of a tendency towards proportional errors has been displayed by Bland-Altman plots (Fig. 4, 7, and 8) with errors equally distributed around the mean difference for all parameters. For temporal parameters, the PE in all the GI phases has been not higher than 14% of the FP value with the lower values for APA1$_d$ and GI$_d$ (Table II). The ICC$_A$ and ICC$_C$ values have pointed out excellent agreement for APA sub-phases duration and good for LOC$_d$ and GI$_d$.

Eventually, linear regression analysis has revealed temporal parameters highly correlated with r values greater than 0.88 (Fig. 5) and the Bland-Altman plots that have presented errors randomly spread around the mean line (Fig. 6).

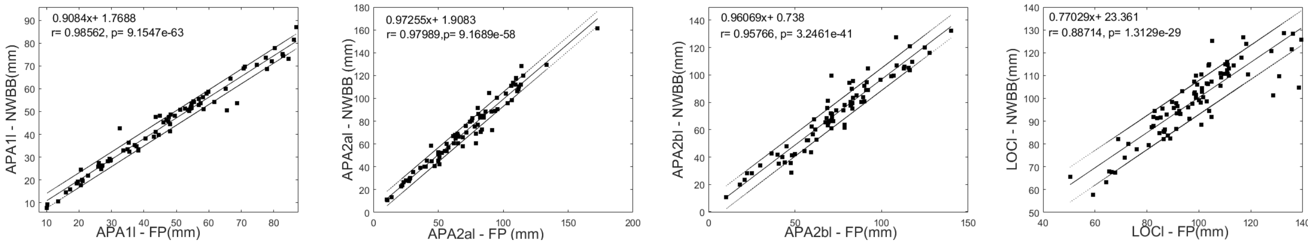


Fig. 3: Spatial parameter for postural (APA1, APA2a, APA2b) and LOC phase: scatter plot and linear regression lines with the regression equation, the Pearson's coefficient and related level of significance.

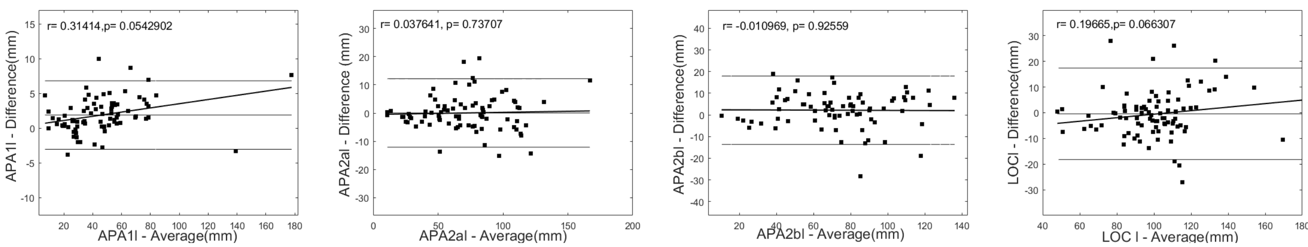


Fig. 4: Spatial parameters: Bland-Altman plots with data randomly spreaded around the mean line. No evidence of proportional errors can be recognized.

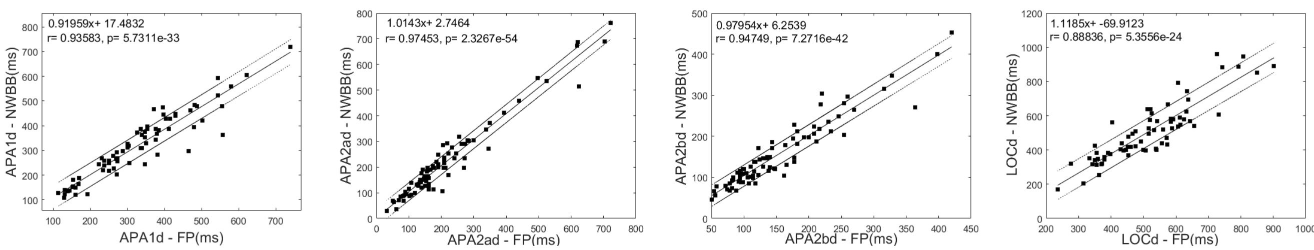


Fig. 5: Temporal parameter for postural (APA1, APA2a, APA2b) and LOC phase: scatter plot and linear regression lines with the regression equation, the Pearson's coefficient and related level of significance.

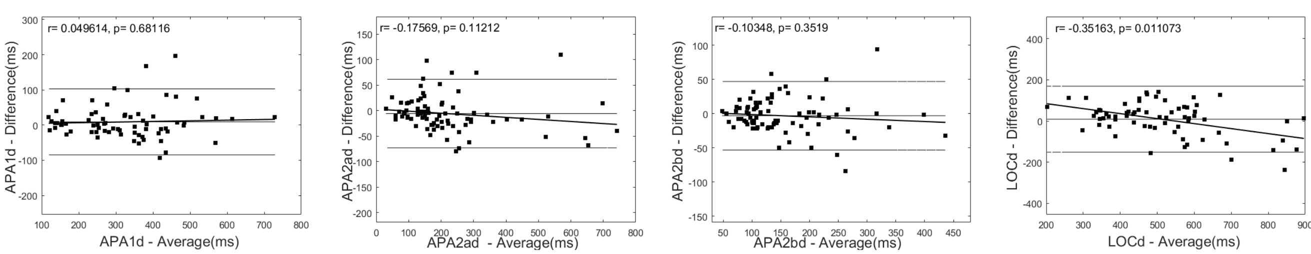


Fig. 6: Temporal parameters for postural (APA1, APA2a, APA2b) and LOC phase: Band-Altman plots with data randomly distributed around the mean line. No evidence of proportional errors can be recognized.

IV. Discussion

The transition from stable bi-pedal to unstable mono-pedal stance during GI requires the body weight to be transferred from one leg to the other, before movement begins. The anticipatory postural adjustments, i.e. APAs, are the way the CNS counteracts the possible loss of equilibrium due to these balance perturbations. Given their relationship with the CNS motor control action, APAs are affected by many factors as the age, the body weight, the presence of neurodegenerative diseases and joint prosthesis or limb amputations [16], [17], [21], [22], [24]–[26]. By providing relevant information about the subject's ability in balance maintenance, APAs allow the early detection of clinical signs, the evaluation of progress of the motor disorders together with a possible prediction of risk of falls [17]. The technological solution commonly adopted for APAs detection lies on the laboratory-grade FP, able to measure the forces exchanged with the ground so that allowing the direct computation of the CoP time-series. With the development of miniaturized inertial sensors, their applicability on body attitude estimation and motion tracking was largely investigated [27], [28]. However only few studies

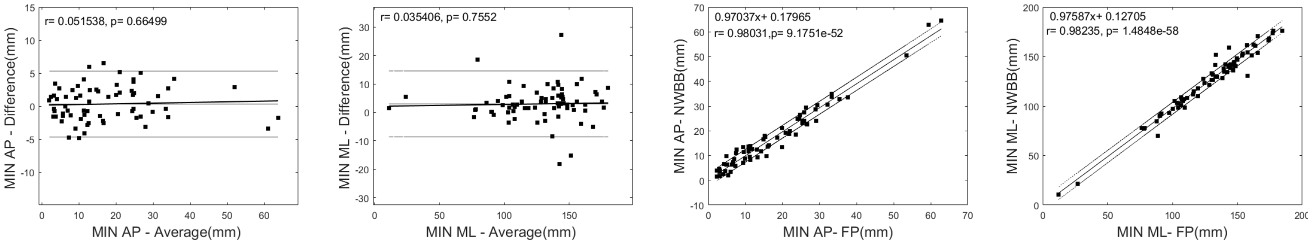


Fig. 7: Spatial parameters (MIN$_{AP}$ and MIN$_{ML}$): Bland-Altman plots with data randomly spreaded around the mean line and no evidence of proportional errors; scatter plot and linear regression line with the regression equation, the Pearson's coefficient and related level of significance are also indicated.

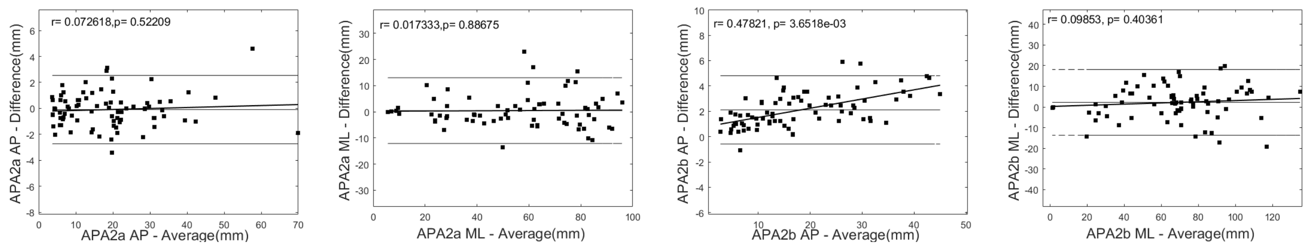


Fig. 8: Spatial parameters (APA2a$_{AP}$, APA2a$_{ML}$, APA2b$_{AP}$ and APA2b$_{ML}$): Band-Altman plots with data randomly spread around the mean line.

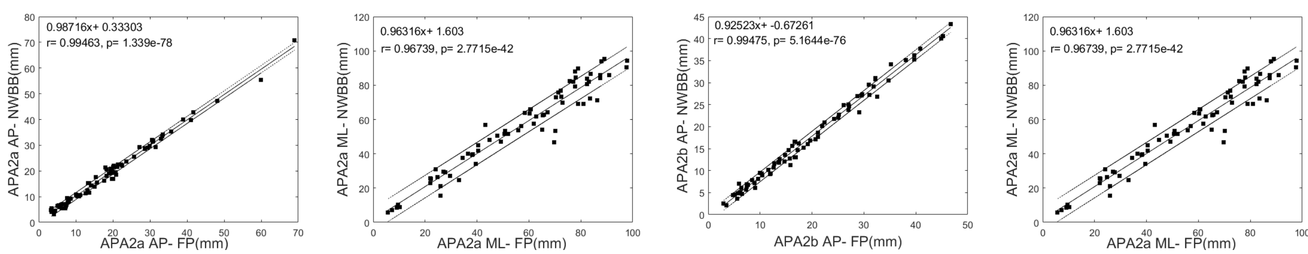


Fig. 9: Spatial parameters (APA2a$_{AP}$, APA2a$_{ML}$, APA2b$_{AP}$ and APA2b$_{ML}$): scatter plot and linear regression line with regression equation, the Pearson's coefficient and related level of significance.

TABLE I: AE and PE (mean $\pm$, standard deviation and 95% CI), intraclass correlation coefficients (ICC$_A$ and ICC$_C$ with their 95% CI) computed for all CoP spatial parameters.

Parameters	CoP$_{FP}$ versus CoP$_{NWBB}$			
	$AE(mm)$	PE	ICC$_A$	ICC$_C$
APA1$_l$	3.2±3.1 (2.5, 3.9)	6.9±6.0 (5.6, 8.2)	0.97 (0.96,0.98)	0.98 (0.98,0.99)
APA2a$_l$	4.5±4.2 (3.6, 5.4)	6.6±5.6 (5.4, 7.8)	0.98 (0.97,0.99)	0.98 (0.97,0.99)
APA2b_l	5.9±6.0 (4.6 7.2)	6.0±5.7 (4.7, 7.2)	0.88 (0.81,0.92)	0.88 (0.82,0.92)
LOC$_l$	6.5±5.2 (5.3, 7.7)	9.6±8.1 (7.8, 11.5)	0.95 (0.93,0.97)	0.96 (0.93,0.97)
MIN$_{AP}$	2.0±1.3 (1.7, 2.4)	15.6±9.3 (12.8, 17.3)	0.97 (0.95,0.98)	0.97 (0.95,0.98)
MIN$_{ML}$	4.8±4.6 (3.7, 4.8)	3.9±2.0 (3.0, 4.8)	0.98 (0.97,0.98)	0.98 (0.97,0.99)
APA2a$_{AP}$	1.1±0.87 (0.8, 1.2)	8.4±6.9 (6.3, 10.6)	0.99 (0.99,0.99)	0.99 (0.99,0.99)
APA2a$_{ML}$	4.7±4.4 (3.6, 5.7)	9.1±8.1 (7.0, 10.9)	0.97 (0.95,0.98	0.97 (0.95,0.98)
APA2b$_{AP}$	6.6±5.2 (5.44, 7.8)	13.6±8.8 (8.3, 14.8)	0.96 (0.93,0.97)	0.96 (0.93,0.97)
APA2b$_{ML}$	2.1±1.3 (1.8, 2.4)	12.8±7.5 (11.1, 14.5)	0.97 (0.96,0.98)	0.99 (0.99,0.99)

TABLE II: AE and PE (mean $\pm$, standard deviation and 95% CI), intraclass correlation coefficients (ICC$_A$ and ICC$_C$ with their 95% CI) computed for CoP temporal parameters.

Parameters	CoP$_{FP}$ versus CoP$_{NWBB}$			
	$AE(ms)$	PE	ICC$_A$	ICC$_C$
APA1$_d$	32±36 (23, 41)	9.7±9.0 (7.6, 11.8)	0.93 (0.889,0.96)	0.94 (0.89,0.96)
APA2a$_d$	25±23 (20, 31)	13.4±10.8 (11.1, 15.8)	0.97 (0.96,0.98)	0.97 (0.96,0.98)
APA2b$_d$	19±17 (15, 23)	13.2±9.3 (11.2, 15.3)	0.95 (0.92,0.96)	0.95 (0.92,0.96)
LOC_d	64±59 (52, 77)	12.7±9.3 (10.4, 14.9)	0.86 (0.796,0.91)	0.86 (0.79,0.91)
GI_d	120±101 (99, 142)	9.9±8.3 (8.2, 11.7)	0.83 (0.75,0.88)	0.84 (0.77,0.89)

examined the possibility to detect events APAs-related by positioning the sensing unit on the trunk or on other body segments, as the shank [21], [29]. Their results showed that some characteristics in the signals of the inertial sensors appeared to be related to two early APA phases, but a direct and complete measurement of them has not yet been reached. Also the use of a number of inertial sensors requires specific technical skills for their placement, thus limiting their usability in a clinical context.

In the clinical practice or in non-high specialized labs, the possibility to adopt a portable and low cost device to obtain CoP trajectory, instead of a more expensive FP, appears a valuable opportunity. The NWBB has been evaluated for static and quasi-static motor tasks, resulting a valid alternative to gold-standard FP and supporting its use in clinical and research applications [1], [7]–[15], [30], [31]. However the adoption of such kind of device requires a task specific validation procedure to conclude that possible differences can be ascribed to real clinically meaningful aspects [9]. In this perspective, Lee and co-workers [22] used NWBB during GI with a pathological and a healthy group of subjects, by finding differences in some CoP parameters between the examined populations. Although their results appear to be promising [22], a limitation of the study, also reported by the same authors, was the lack of the analysis of accuracy and precision, required before the clinical use. To the best of author knowledge, no studies compared the performances of the NWBB with respect to laboratory-grade FP during GI. However, the assessment of the concurrent validity, i.e. the comparison with a FP with simultaneous measurements, is of fundamental importance in order to clarify if the errors could limit its usability for this motor task, widely used for clinical investigations [19].

According to [9], [14], findings of the present study have highlighted a high similarity between CoP trajectories from both the devices, strengthened by the low values of mean RMSE in AP and ML directions for each GI phase (Fig. 2). Specifically, the RMSE values on the whole trajectory (8.3$\pm$5.2 mm for the AP and 12.4$\pm$6.9 mm for the ML) are in line with the results of Eguchi and co-workers, obtained during gait [31]. In particular, in the AP direction, a RMSE lower than 3 mm has been observed for all the APA phases (Fig. 2) suggesting the suitability of the NWBB by preserving the information content of each the GI phases [16], [24]. Incidentally, the lower accuracy in ML-CoP component for all the GI phases (Fig. 2) confirms what shown during quite standing by Leach and co-workers [32] for which the RMSE was significantly greater in ML than in AP direction.

Moreover, in the present study, the postural phases appear characterized by lower errors than in the dynamic one (LOC), for which the RMSE is around 7 mm in both the CoP components (Fig. 2). This finding related to the LOC phase are comparable with data obtained by Egouchi et al. [31] during walking straight for which the mean RMSE values have been of 9.7 mm in AP and 6.2 mm in ML direction. During the LOC phase, the CoP displaces forward to anticipate and control the center of mass movement during the last part of the stance. With respect to the APAs, this phase is characterized by a large and faster CoP displacement and it is likely to assume that the lack of horizontal forces for the NWBB can mostly affect this spatial parameter. This hypothesis is supported by the regression analysis performed on AE values during LOC phase with respect to the displacement of CoP obtained by the FP data: as the sway increases, the related error increases, as indicated by the positive slope of the linear trend line and by the significant correlation (Fig. 10(a)). In particular, it appears that the main contribution of the measurement error, in LOC phase, is given by the absence of horizontal force components, characterized by a peak equal to 27.3 N (on average) in AP and 15.0 N in ML direction. This aspect has been highlighted by the linear regression analysis: the higher is the amplitude of AP-GRF component, the higher is the AE (Fig. 10(b)).

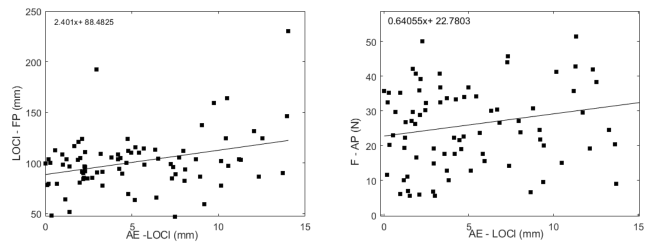


Fig. 10: Scatter plots and linear regression analysis for the AE values in the LOC phase with respect to FP data for the LOC length (a, r=0.64, p=0.02) and the AP component of GRF (b, r=0.51, p=0.047).

The remarkable correspondence between NWBB and FP trajectories has been pointed out by the RMSE values and also by the limited PE and AE relative to the APA and LOC lengths, as well as by the ICC values that have revealed consistency and a general excellent agreement between measures (Table I). Specifically, length measures have been characterized by low percentage errors, always under 10%. APA parameters in AP and ML direction (MIN$_{AP,ML}$, APA2a$_{AP,ML}$ and APA2b$_{AP,ML}$, Table I) have appeared more variables, with PE ranging from 3.9% to 15.6%, which mirrors in the worst case an AE not higher than 6.6 mm. Taking into account the previous findings, it should be noticed that, the NWBB has preserved the ratios between each postural and locomotor phase length and the total path length with respect to FP data. Specifically, the mutual relations are equal to 16% for APA1$_l$ (16%$\pm$6 in FP and 16%$\pm$9 in NWBB), around 24% for both the APA2 phase (24%$\pm$1 in FP and 25%$\pm$10 in NWBB) and finally not higher than 36% for the LOC (34%$\pm$6 in FP and 35%$\pm$6 in NWBB). This finding encourages the use of NWBB in clinical settings or for follow-up studies, when the phases length variations have to be monitored to evaluate the effects of the motor control deterioration or to discriminate pathology from physiological behavior [16], [18], [21], [22], [25].

To support this consideration, it must also be said that unlikely the errors would produce wrong evaluations of the main spatial parameters of GI, given the high consistency and significant linearity of the NWBB measures (Fig. 7 and 9) with a Pearson correlation coefficient (r) always higher than 0.88. This is also in line with what observed in a in quasi-static tasks by Mengarelli and coworkers [14]. Furthermore, the substantial accuracy of NWBB measures has been confirmed by the excellent values of ICC agreement and consistency (Table I) and by the absence of tendency toward proportional errors observed in Bland-Altman plots (Fig. 7 and 8) where values are equally distributed around the mean difference line. Though these features indicated the not complete fulfillment of all the interchangeability conditions, as reported in [9], [14], [33], nevertheless this does not compromise the use of NWBB for GI analysis in a single study context, when the same device is adopted to perform measurements.

Note that in this study multiple trials from the same subjects were considered as independent. This aligns with other previous studies [7] and it appears supported by considering that the average SD of CoP displacement among all 100 trials resulted comparable with that computed separately for each subject: 14.13 mm vs. 12.14 mm for AP direction and 48.18 mm vs. 44.33 mm for ML direction. Thus, it appears reasonable to consider multiple trials from a single subject as not repeatable or biased, since a single subject exhibits a postural response variability comparable with respect to that observed if multiple subjects are considered.

Finally, a new relevant aspect not accurately investigated in other similar studies, has been related to the reliability of NWBB measures for time parameters. NWBB was used by Lee and coworkers [22] in GI test by finding fair within-subject reliability ($ICC = 0.517$) for CoP path time and by Eguchi [31] in gait by obtaining for stance duration an accuracy equivalent to a foot switch system. Results of the present study (Table II) have shown significant linear correlation between temporal parameters obtained from both the devices (Fig. 5, $r \geq 0.88$) and PE between 9% and 14% (Table II). In addition, ICC_A and ICC_C values have shown good (≥ 0.83) agreement or longer GI phase and excellent (I$\geq$ 0.93) shorter phases.

V. Conclusion

The NWBB use has been documented in a number of scientific papers in which GRF and CoP have been analyzed in healthy and pathological subjects, to describe posture and a variety of functional tasks having clinical applications [1], [2], [8], [11], [15], [20], [31], [34]–[37]. The present study pursues the intent to enhance NWBB validity by widening the set of possible motor tasks that can be investigated with this device. In this perspective, the present work fills a gap providing relevant information about the validity of NWBB in APAs analysis.

The limited errors with fixed biases and high consistency achieved in this study, show the suitability of NWBB in quantifying the time and spatial parameters in all the GI phases. As observed in [7], [8], [14], the mainly contribution of errors is due to the lack of information given by the absence of the horizontal components of GRF. According to [38], a second source of inaccuracy can be ascribed to the technological limits of the the NWBB sensors, that determine percentages errors in the vertical component of GRF thus affecting the derived parameters. However such sources of inaccuracy, device-dependent, do not affect the validity of NWBB in monitoring APAs, given the limited ranges of the errors values. Future studies should be devoted specifically to disease-focused evaluations of the NWBB suitability for GI investigation in those pathologies affecting balance capabilities. In addition, they could verify the inter-device and test-retest reliability during a GI test, to completely state its the suitability for clinical applications. A possible further investigation could be focused on the use of specific procedure, as the linear calibration proposed by [32] to obtain an errors reduction. It should however noticed that this kind of procedure is possible if FP data are available.

References

[1] G. Severini, S. Straudi, C. Pavarelli, M. Da Roit, C. Martinuzzi, L. D. M. Pizzongolo, and N. Basaglia, "Use of nintendo wii balance board for posturographic analysis of multiple sclerosis patients with minimal balance impairment," *Journal of Neuroengineering and Rehabilitation*, vol. 14, no. 1, p. 19, 2017.

[2] K. Otte, B. Kayser, S. Mansow-Model, J. Verrel, F. Paul, A. U. Brandt, and T. Schmitz-Hübsch, "Accuracy and reliability of the kinect version 2 for clinical measurement of motor function," *PloS One*, vol. 11, no. 11, p. e0166532, 2016.

[3] Y. Jiang, V. Hernandez, G. Venture, D. Kulić, and B. K Chen, "A data-driven approach to predict fatigue in exercise based on motion data from wearable sensors or force plate," *Sensors*, vol. 21, no. 4, p. 1499, 2021.

[4] L.-F. Shi, H. Liu, G.-X. Liu, and F. Zheng, "Body topology recognition and gait detection algorithms with nine-axial immu," *IEEE Transactions on Instrumentation and Measurement*, vol. 69, no. 3, pp. 721–728, 2019.

[5] T. Liu, Y. Inoue, and K. Shibata, "A wearable force plate system for the continuous measurement of triaxial ground reaction force in biomechanical applications," *Measurement Science and Technology*, vol. 21, no. 8, p. 085804, 2010.

[6] F. Sgrò, G. Monteleone, M. Pavone, and M. Lipoma, "Validity analysis of wii balance board versus baropodometer platform using an open custom integrated application," *AASRI Procedia*, vol. 8, pp. 22–29, 2014.

[7] A. Huurnink, D. P. Fransz, I. Kingma, and J. H. van Dieën, "Comparison of a laboratory grade force platform with a nintendo wii balance board on measurement of postural control in single-leg stance balance tasks," *Journal of Biomechanics*, vol. 46, no. 7, pp. 1392–1395, 2013.

[8] R. A. Clark, B. F. Mentiplay, Y.-H. Pua, and K. J. Bower, "Reliability and validity of the wii balance board for assessment of standing balance: a systematic review," *Gait & Posture*, vol. 61, pp. 40–54, 2018.

[9] A. Mengarelli, S. Cardarelli, A. Strazza, F. Di Nardo, S. Fioretti, and F. Verdini, "Validity of the nintendo wii balance board for the assessment of balance measures in the functional reach test," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 26, no. 7, pp. 1400–1406, 2018.

[10] P. Scaglioni-Solano and L. F. Aragón-Vargas, "Validity and reliability of the nintendo wii balance board to assess standing balance and sensory integration in highly functional older adults," *International Journal of Rehabilitation Research*, vol. 37, no. 2, pp. 138–143, 2014.

[11] B. Bonnechere, O. Van Hove, B. Jansen, and S. V. S. Jan, "Validation of the wii balance board to assess static balance during dual-task activity in healthy subjects," *Medicine in Novel Technology and Devices*, vol. 1, p. 100003, 2019.

[12] L. Castelli, L. Stocchi, M. Patrignani, G. Sellitto, M. Giuliani, and L. Prosperini, "We-measure: Toward a low-cost portable posturography for patients with multiple sclerosis using the commercial wii balance board," *Journal of the Neurological Sciences*, vol. 359, no. 1-2, pp. 440–444, 2015.

[13] N. Tweed, S. Williams, D. Williams, and J. Dingley, "Utility of a low-cost wireless force platform as a potential clinical test of balance recovery after neuraxial anaesthesia," *International Journal of Obstetric Anesthesia*, vol. 23, no. 3, pp. 227–232, 2014.

[14] A. Mengarelli, F. Verdini, S. Cardarelli, F. Di Nardo, L. Burattini, and S. Fioretti, "Balance assessment during squatting exercise: A comparison between laboratory grade force plate and a commercial, low-cost device," *Journal of Biomechanics*, vol. 71, pp. 264–270, 2018.

[15] G. Yamako, E. Chosa, K. Totoribe, Y. Fukao, and G. Deng, "Quantification of the sit-to-stand movement for monitoring age-related motor deterioration using the nintendo wii balance board," *PLoS One*, vol. 12, no. 11, 2017.

[16] N. Cau, V. Cimolin, M. Galli, H. Precilios, E. Tacchini, C. Santovito, and P. Capodaglio, "Center of pressure displacements during gait initiation in individuals with obesity," *Journal of Neuroengineering and Rehabilitation*, vol. 11, no. 1, p. 82, 2014.

[17] R. Martinez-Mendez, M. Sekine, and T. Tamura, "Detection of anticipatory postural adjustments prior to gait initiation using inertial wearable sensors," *Journal of Neuroengineering and Rehabilitation*, vol. 8, no. 1, p. 17, 2011.

[18] A. Delval, K. Dujardin, C. Tard, H. Devanne, S. Willart, J.-L. Bourriez, P. Derambure, and L. Defebvre, "Anticipatory postural adjustments during step initiation: elicitation by auditory stimulation of differing intensities," *Neuroscience*, vol. 219, pp. 166–174, 2012.

[19] J. G. Remelius, J. Hamill, J. Kent-Braun, and R. E. Van Emmerik, “Gait initiation in multiple sclerosis,” *Motor Control*, vol. 12, no. 2, pp. 93–108, 2008.

[20] R. Zhong and P.-L. P. Rau, “Are cost-effective technologies feasible to measure gait in older adults? a systematic review of evidence-based literature,” *Archives of Gerontology and Geriatrics*, vol. 87, p. 103970, 2020.

[21] M. Mancini, C. Zampieri, P. Carlson-Kuhta, L. Chiari, and F. B. Horak, “Anticipatory postural adjustments prior to step initiation are hypometric in untreated parkinson’s disease: an accelerometer-based approach,” *European Journal of Neurology*, vol. 16, no. 9, pp. 1028–1034, 2009.

[22] J. Lee, G. Webb, A. P. Shortland, R. Edwards, C. Wilce, and G. D. Jones, “Reliability and feasibility of gait initiation centre-of-pressure excursions using a wii® balance board in older adults at risk of falling,” *Aging Clinical and Experimental Research*, vol. 31, no. 2, pp. 257–263, 2019.

[23] H. L. Bartlett, L. H. Ting, and J. T. Bingham, “Accuracy of force and center of pressure measures of the wii balance board,” *Gait & Posture*, vol. 39, no. 1, pp. 224–228, 2014.

[24] E. Yiou, T. Caderby, A. Delafontaine, P. Fourcade, and J.-L. Honeine, “Balance control during gait initiation: State-of-the-art and research perspectives,” *World Journal of Orthopedics*, vol. 8, no. 11, p. 815, 2017.

[25] A. Delval, P. Krystkowiak, J.-L. Blatt, E. Labyt, J.-L. Bourriez, K. Dujardin, A. Destée, P. Derambure, and L. Defebvre, “A biomechanical study of gait initiation in huntington’s disease,” *Gait & Posture*, vol. 25, no. 2, pp. 279–288, 2007.

[26] W. Young, S. Ferguson, S. Brault, and C. Craig, “Assessing and training standing balance in older adults: a novel approach using the ‘nintendo wii’balance board,” *Gait & Posture*, vol. 33, no. 2, pp. 303–305, 2011.

[27] S. Díaz, J. B. Stephenson, and M. A. Labrador, “Use of wearable sensor technology in gait, balance, and range of motion analysis,” *Applied Sciences*, vol. 10, no. 1, p. 234, 2020.

[28] A. Mengarelli, S. Cardarelli, A. Tigrini, L. Marchesini, A. Strazza, S. Fioretti, and F. Verdini, “Inertial sensor based estimation of the center of pressure during unperturbed upright stance,” in *2019 IEEE 23rd International Symposium on Consumer Technologies (ISCT)*. IEEE, 2019, pp. 186–191.

[29] G. Bonora, M. Mancini, I. Carpinella, L. Chiari, F. B. Horak, and M. Ferrarin, “Gait initiation is impaired in subjects with parkinson’s disease in the off state: Evidence from the analysis of the anticipatory postural adjustments through wearable inertial sensors,” *Gait & Posture*, vol. 51, pp. 218–221, 2017.

[30] T. B. Weaver, C. Ma, and A. C. Laing, “Use of the nintendo wii balance board for studying standing static balance control: technical considerations, force-plate congruency, and the effect of battery life,” *Journal of Applied Biomechanics*, vol. 33, no. 1, pp. 48–55, 2017.

[31] R. Eguchi and M. Takahashi, “Validity of the nintendo wii balance board for kinetic gait analysis,” *Applied Sciences*, vol. 8, no. 2, p. 285, 2018.

[32] J. M. Leach, M. Mancini, R. J. Peterka, T. L. Hayes, and F. B. Horak, “Validating and calibrating the nintendo wii balance board to derive reliable center of pressure measures,” *Sensors*, vol. 14, no. 10, pp. 18 244–18 267, 2014.

[33] J. Lee, D. Koh, and C. Ong, “Statistical evaluation of agreement between two methods for measuring a quantitative variable,” *Computers in Biology and Medicine*, vol. 19, no. 1, pp. 61–70, 1989.

[34] S. Verma, D. Kumar, A. Kumawat, A. Dutta, and U. Lahiri, “A low-cost adaptive balance training platform for stroke patients: a usability study,” *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 25, no. 7, pp. 935–944, 2017.

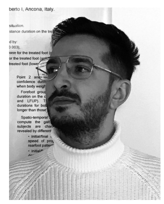
[35] S. Abujaber, G. Gillispie, A. Marmon, and J. Zeni Jr, “Validity of the nintendo wii balance board to assess weight bearing asymmetry during sit-to-stand and return-to-sit task,” *Gait & Posture*, vol. 41, no. 2, pp. 676–682, 2015.

[36] J. Y. Lim, Y. Yi, S. W. Jung, and D.-S. Park, “Comparison of vertical ground reaction forces between female elderly and young adults during sit-to-stand and gait using the nintendo wii balance board,” *Physical Therapy Rehabilitation Science*, vol. 7, no. 4, pp. 179–185, 2018.

[37] J. Ruff, T. L. Wang, C. C. Quatman-Yates, L. S. Phieffer, and C. E. Quatman, “Commercially available gaming systems as clinical assessment tools to improve value in the orthopaedic setting: a systematic review,” *Injury*, vol. 46, no. 2, pp. 178–183, 2015.

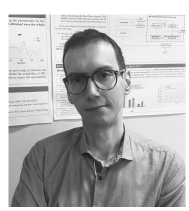
[38] F. Verdini, A. Mengarelli, A. Tigrini, A. Strazza, F. Di Nardo, S. Cardarelli, and S. Fioretti, “Accuracy evaluation of force measurement through the wii balance board during squat and functional reach tests,” in *2019 IEEE 23rd International Symposium on Consumer Technologies (ISCT)*. IEEE, 2019, pp. 291–295.

Federica Verdini obtained the MS degree in Electronic Engineering at the Ancona University in 1997 and the Ph.D degree in Bioengineering at the Bologna University in 2001. Her research activity focuses on the recording and processing of biological signals associated with human movement and on the study of control models related to motion and posture. In these fields she authored many scientific publications in international journals, books and conference proceedings. Currently she is Chief-staff scientist at the Movement analysis and Bioengineering Laboratory of the Università Politecnica delle Marche.



Andrea Tigrini received the MS degree in Biomedical Engineering from Università Politecnica delle Marche, Ancona, Italy, in 2018. Currently, he is a Ph.D. student in Information Engineering at Università Politecnica delle Marche. His research interests involve the study of human biomechanics, neuromuscular control and rehabilitation engineering. In these fields, he authored/coauthored scientific publications in international journals and conference proceedings.

Sandro Fioretti graduated in 1979 in Electronic Engineering at Ancona University and currently is Associate Professor in Bioengineering at the Department of Information Engineering, Università Politecnica delle Marche. His main research interests are in the field of human movement analysis and its related fields such as stereophotogrammetry, linear and nonlinear filtering, joint kinematics, static and perturbed posturography, gait analysis. He is the author of numerous scientific publications in international journals, books and congress proceedings.



Alessandro Mengarelli received the MS degree in Biomedical Engineering and the Ph.D. degree in Information Engineering from Università Politecnica delle Marche, in 2012 and 2017, respectively. Currently, his research interests involve biomechanics of movement in both static and dynamic conditions, the application of control theory and modeling to the analysis of balance and the non-linear analysis of timeseries. Further, he is also involved in the assessment of muscular functions during gait and in myoelectric pattern recognition. In these fields, he authored/coauthored scientific publications in peer reviewed international journals and conference proceedings.