EXPERIMENTAL VALIDATION OF A TORSO-MOUNTED INERTIAL SENSOR SYSTEM FOR HUMAN BALANCE AND MOTION ASSESSMENT

Marko Čeredar[1*] – Aleksandar Sušić[1] – Bruno Dogančić[1] – Martina Lovrenić-Jugović[2]

[1]Department of technical mechanics, Faculty of Mechanical Engineering and Naval Architecture, University of Zagreb, Ivana Lučića 5, Zagreb, Croatia
[2]Faculty of Metallurgy, University of Zagreb, Aleja narodnih heroja 3, Sisak, Croatia

ARTICLE INFO

***Article history*:**
Received: 6. 4. 2026.
Received in revised form: 27. 4. 2026.
Accepted: 18. 7. 2026.

Keywords:
Centre of mass estimation
Ergonomic assessment
Industry 4.0
Motion validation
Wearable inertial sensors (IMU)

DOI: 10.30765/er.3406

Abstract:

Wearable inertial measurement units (IMUs) provide a mobile and cost-effective alternative to laboratory motion analysis and are increasingly used in Industry 4.0 for ergonomic assessment, workload evaluation, and biomimetic modelling in human–robot interaction. However, the validity of IMU-based body centre of mass (CoM) estimation depends on sensor configuration. This pilot technical validation study evaluates agreement between an upper-torso- and upper-extremities mounted IMU and a pedobarographic platform during quasi-static standing with controlled torso rotation. Lateral and sagittal CoM displacements were compared using RMSE, correlation coefficient, area-under-curve (AUC) difference, and frequency-domain analysis. Higher agreement was observed in the left–right direction (RMSE = 1.514 cm, r = 0.851), suggesting that the partial IMU setup may be suitable for monitoring lateral torso sway relevant to balance and postural load under quasi-static conditions. In the forward–backward direction, agreement was low (r = 0.488) due to the absence of lower-limb motion capture. Results demonstrate both the potential and limitations of partial IMU setups for ergonomic monitoring and task optimization in industrial environments.

ACCEPTED

1 Introduction

Industry 4.0 has greatly impacted the integration of monitoring technologies into industrial settings by introducing smart process optimization and connected, data-driven production systems. This has led to the use of portable sensory equipment for motion analysis and ergonomic evaluation. Within smart manufacturing frameworks, wearable sensing technologies are being applied for workload monitoring, injury prevention, and adaptive human-robot collaboration, as ergonomic risk assessment has become essential for reducing musculoskeletal disorders and improving worker safety in industrial assembly systems [1] - [4]. For example, Naranjo et al. [2] pointed out the role of wearable sensory equipment in industrial environments, while Baklouti et al. [3] showed the application of IMU-based systems in ergonomic evaluations while performing work tasks in an industrial setting.

When such applications are taken into account, biomechanical parameters are of great value, out of which the centre of mass (CoM) is of particular importance. CoM movement and displacement are often used to assess postural stability, movement efficiency, and balance control, which are all linked with risk of injury as well as workplace safety. With that in mind, CoM is used in ergonomic research as well as fall risk assessment, as deviations beyond the limits of stability are linked to loss of balance and increased fall incidence in workplace conditions [5] - [7]. The importance of CoM dynamics has been described as an inverted pendulum, where still quiet standing balance is achieved through distribution of contact pressure between the feet and the ground, maintaining and adjusting the centre of pressure (CoP), as stated in postural control models [8] - [10].

* Corresponding author
E-mail address: marko.ceredar@fsb.unizg.hr

Traditionally, CoM estimation and movement analysis have relied on laboratory-based optical motion capture (MoCap) systems and force platforms, which provide high spatial and temporal accuracy and are widely considered reference standards in biomechanics research [11], [12]. Optical systems, such as VICON, compute segment trajectories by tracking markers placed on characteristic landmarks utilizing triangulation [12]. As for force plates, they measure ground reaction forces and determine the CoP as a weighted centroid of the forces applied on the plate's surface [6], [13]. While both of those systems provide highly precise measurements, these laboratory systems are limited in industrial settings due to high cost, limited portability, as well as controlled environments [11].

To tackle the limitations of laboratory-based MoCap, inertial measurement units (IMUs) have emerged as an alternative system for motion capture [14] - [16]. IMUs consist of accelerometers and gyroscopes, which capture kinematic data. Such data are then processed and, through biomechanical modelling, are used to estimate segmental kinematics from which parameters such as CoM movement can be obtained [16] - [18]. Ferrari et al. [14] proved the validity of trunk-mounted IMUs for postural assessment, while Naomani et al. [15] stated that there was acceptable agreement between force platform measurements in static conditions and IMU data. With the advancement of sensor fusion, filtering, and synchronization, IMU-based estimation accuracy has greatly improved through noise reduction and segmental orientation computation [17], [18].

However, even with all the improvements, the accuracy of IMU-measured data and CoM estimation still depends on sensor placement, biomechanical models used, as well as the number of segments tracked [16], [18]. In an industrial setting, whole-body tracking with IMUs is often impractical due to long preparation and calibration times, work task execution practicality, as well as worker comfort [19], [20]. With that said, simplified IMU configurations with fewer sensors, such as isolated torso-mounted systems, have been explored as a faster solution for large-scale ergonomic monitoring [21], [22]. Shepherd and Giblin [23] have demonstrated the effectiveness of simplified IMU setups in performance monitoring, while Kongsawasdi et al. [24] showed the practicality of IMU-based postural sway in quiet stance assessment. The main limiting factor of upper-body IMU systems comes from the fact that reducing sensor coverage limits the accuracy of assessing whole-body dynamics, particularly the effect of the lower-extremity contribution [16].

It is important to note that IMU systems and pedobarographic platforms, which are used in this study, are not equal in their data capture methods, and as such, they do not quantify identical physical variables. IMUs generate body kinematics and estimate CoM position [16] - [18], whereas pedobarographic platforms measure plantar pressure distribution and determine CoP as a weighted centroid derived from ground reaction forces [6], [13]. Under quasi-static standing conditions, the CoP approximates the vertical projection of the CoM, forming the biomechanical basis for comparative analysis [8] - [10]. Studies involving simultaneous measurement of CoP and trunk or whole-body motion have investigated these variables as complementary indicators of postural regulation and balance control [25]. Nevertheless, CoP represents the control output of postural regulation, while CoM reflects the mechanical state of body mass distribution, and their relationship becomes more complex during dynamic movements [9], [10].

Most validation studies to date have compared IMU measurements with optical MoCap systems, thereby evaluating similar kinematic outputs [11], [14], [15], and a few studies have examined the agreement between IMU-derived CoM displacement and CoP measures obtained from force plates [15], [16]. In contrast, the present study evaluates agreement between model-based IMU-derived CoM displacement and pressure-derived CoP displacement obtained from a pedobarographic platform. This provides a distinct validation perspective because the compared systems are biomechanically related but based on different physical measurement principles, while both are more suitable for field deployment than laboratory optical motion capture.

Based on this rationale, the purpose of the present study was to evaluate the agreement between IMU-derived CoM displacement and pedobarographic-platform-derived CoP displacement during controlled trunk motion under quasi-static standing conditions.

2 Methods and subjects

2.1 Experimental Setup

To validate the measurement system, a wearable IMU system was used and compared with a pedobarographic platform as the reference measurement system. The pedobarographic platform used in the study was a Tekscan platform, model MobileMat, with sensel area of 48.7 x 44.7 cm and a capture frequency

ACCEPTED

of 100 Hz. Tekscan pressure-mapping systems have previously been validated for accurate plantar pressure distribution measurement in biomechanical applications [26]. The pedobarographic platform was selected because under quasi-static standing conditions, CoP approximates the projection of CoM onto the support surface, enabling comparative analysis.

As the wearable inertial measurement system, the Perception Neuron 2.0 system (Noitom Ltd.) was used. It consists of wireless inertial measurement units, which are attached via straps, integrating tri-axial accelerometers and gyroscopes, capable of motion capture at sampling frequencies of up to 120 Hz depending on the selected configuration. The Perception Neuron system has previously been validated for full-body motion capture, demonstrating acceptable agreement with optical reference systems [27]. In the present study, the sensors were applied in a partial configuration positioned on the upper part of the subject's torso and upper extremities, specifically a 9-sensor setup, allowing estimation of trunk kinematics required for the calculation of body centre of mass displacement. Such a configuration was selected to simulate a simplified sensor setup suitable for rapid implementation in industrial conditions, where the application of full-body sensor networks is often not practical. Prior to measurement, the IMU system was calibrated according to manufacturer procedures using a static calibration pose to align sensor orientation with the biomechanical model.

2.2 Subjects and Experimental Protocol

Measurements were conducted on one healthy male subject aged 27 years, with a height of 178 cm and a body mass of 67 kg. The study was designed as a pilot validation involving a single subject to evaluate measurement accuracy prior to larger-scale testing. With that in mind, the validation presented here should be interpreted as a preliminary technical agreement assessment rather than full population-level validation. The experimental protocol consisted of controlled rotational movements of the torso to the left and right while standing on the pedobarographic platform with the partial IMU setup equipped. The experiment was designed to produce quasi-static motion conditions with slow trunk movements and a constant base of support, under which CoP can serve as an approximate proxy for horizontal CoM displacement. In order to ensure movement repeatability and reduce variability caused by uncontrolled arm movements, as well as to keep all the movement in the transverse plane for simplification, the subject held a light, long hollow rod during the execution of the movements as a support, which had a minimal influence on the overall body mass and inertia. During task execution, which consisted of multiple trials of approximately 20 s in length, both measurement systems simultaneously collected data that were later used for the comparative analysis of body centre of mass motion. The experimental setup can be seen in Figure 1 where red dots represent marker placement locations, the pedobarographic platform is blue, and the rod is green.

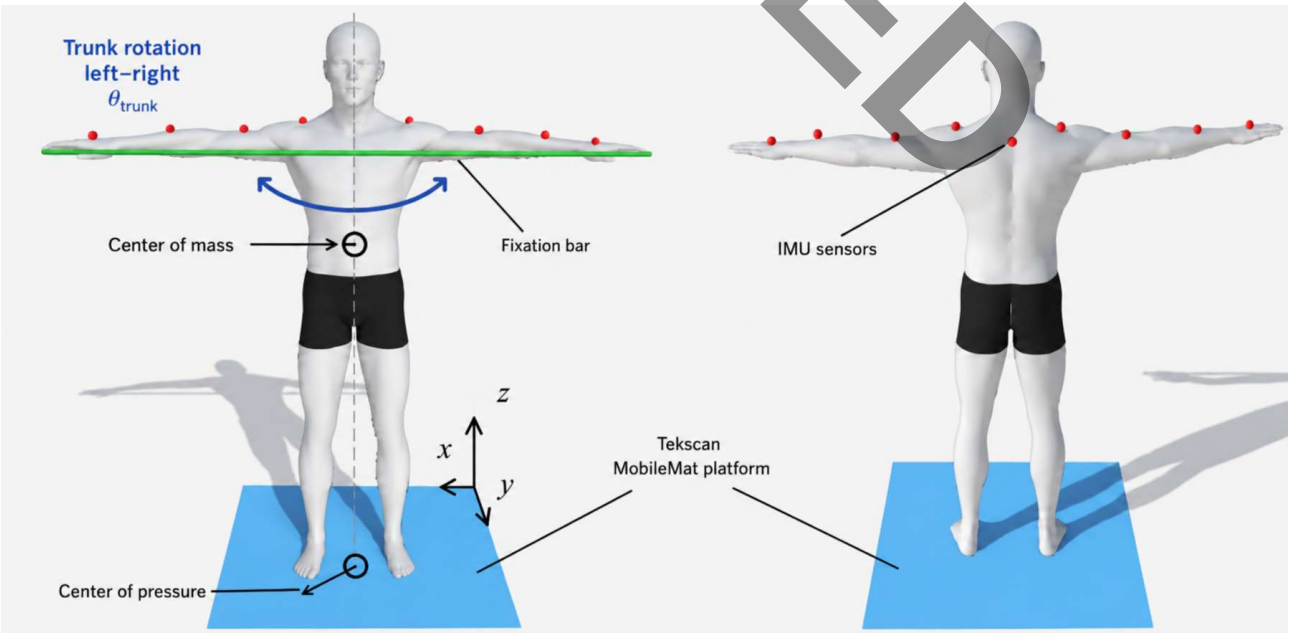


Figure 1. Experimental setup

ACCEPTED

2.3 Data Processing

Raw measurement data from the pedobarographic platform were imported into the MATLAB software package [28], in which CoP movement was calculated as the pressure-weighted centroid of the pressure distribution across the sensor matrix. As for IMU data, kinematic data stored in .bvh format were imported into Biomechanics of Bodies software, which estimated dynamic data via inverse dynamics. After extracting CoM data as a .csv and importing it in MATLAB, all subsequent processing, analyses, and calculations were performed. The data were first temporally synchronized, meaning that the initial recording time points were equalized to enable direct comparison of the signals. Since the IMU system and pedobarographic platform have different sampling frequencies, the signals were aligned by manually identifying starting points and resampling the IMU-derived CoM signal via linear interpolation in MATLAB to match the number of samples and temporal resolution of the pedobarographic platform CoP signal. This ensured that paired data points corresponded to the same relative time instants, which enabled direct point-by-point comparison and application of error metrics. Following resampling, both signals contained an equal number of samples with matched temporal indices, allowing the use of paired-sample statistical metrics as defined in Equations (2)–(6).

After synchronization and extraction of the common time interval, curves of body centre of mass displacement in the lateral (left–right) and sagittal (forward–backward) directions were obtained. The coordinate system was defined with the x-axis representing lateral (left–right) displacement and the y-axis representing sagittal (forward–backward) displacement. A comparative analysis was performed on the obtained curves using the error indicators, root mean square error (RMSE), the correlation coefficient, and Bland–Altman agreement analysis. Additionally, error plots over time were generated, the difference in area-under-curves was calculated as a measure of total signal deviation, and a comparison of the signals in the frequency domain was performed to assess the agreement of the dominant frequency components of body centre of mass motion.

CoP data from pedobarographic platform was calculated as

$$CoP = \frac{\sum_{i=1}^{N} p_i \, x_i}{\sum_{i=1}^{N} p_i} \tag{1}$$

where p_i represents the pressure measured at sensor element i, and x_i denotes the spatial coordinates of each sensel along the *x*-axis.

The root mean square error (RMSE) between CoP data from pedobarographic platform and CoM data from IMU system was calculated as

$$RMSE = \sqrt{\frac{1}{N}\sum_{i=1}^{N}\left(x_{p,i} - x_{m,i}\right)^2} \tag{2}$$

where $x_{p,i}$ and $x_{m,i}$ represent the platform's CoP and IMU's CoM data in lateral direction, while N denotes the number of synchronized samples used in the comparison interval.

The data were additionally evaluated using Pearson's correlation coefficient

$$r = \frac{\sum_{i=1}^{N}\left(x_{p,i} - \bar{x}_p\right)\left(x_{m,i} - \bar{x}_m\right)}{\sqrt{\sum_{i=1}^{N}\left(x_{p,i} - \bar{x}_p\right)^2 \sum_{i=1}^{N}\left(x_{m,i} - \bar{x}_m\right)^2}} \tag{3}$$

where $\bar{x}_p$ and $\bar{x}_m$ represent mean values of corresponding data.

Area-under-curve difference was calculated as

ACCEPTED

$$AUC_{diff} = \sum_{i=1}^{N} |x_{p,i} - x_{m,i}| \Delta t \tag{4}$$

where Δt denotes the sampling interval.
Agreement between the spectral characteristics of the signals was computed by cumulative absolute difference between their normalized Fourier magnitude spectra (normalized by the number of samples)

$$D_{FFT} = \sum_{k=1}^{\frac{N}{2}} |X_p(k) - X_m(k)| \tag{5}$$

where $X_p(k)$ and $X_m(k)$ represent normalized discrete Fourier magnitude spectra.

Finally, Bland–Altman analysis was performed to assess agreement between measurement systems, with limits of agreement defined as

$$LOA = \bar{d} \pm 1.96\sigma_d \tag{6}$$

where $\bar{d}$ is mean difference between paired measurements and σ_d is standard deviation of the differences.
All metrics were computed independently for the lateral (x) and sagittal (y) displacement components.

ACCEPTED

3 Results

3.1 Time-domain comparison of platform-derived CoP and IMU-derived CoM displacement

The comparison of platform-derived CoP and IMU-derived CoM displacement signals is shown in Figure 2. In the lateral (left–right) direction, both systems exhibited similar signal shapes and temporal evolution, with comparable peak locations and oscillation patterns. In the sagittal (forward–backward) direction, the observed differences in amplitude and temporal behaviour were visibly larger. These observations are supported by RMSE and correlation analyses, where the left–right correspondence between the systems was stronger (RMSE = 1.51 cm, r = 0.85) compared with the forward–backward direction (RMSE = 1.53 cm, r = 0.49). The obtained error magnitudes indicate that the torso-mounted IMU system was capable of capturing the principal characteristics of lateral trunk motion, while reduced correspondence was observed for sagittal displacement. The peak-to-peak displacement amplitude during the trials was approximately 7.83 cm in the lateral direction and 9.36 cm in the sagittal direction, indicating that the RMSE values correspond to approximately 19.28% and 16.35% of total displacement range, respectively. These error magnitudes therefore represent a limited proportion of the total displacement amplitude.

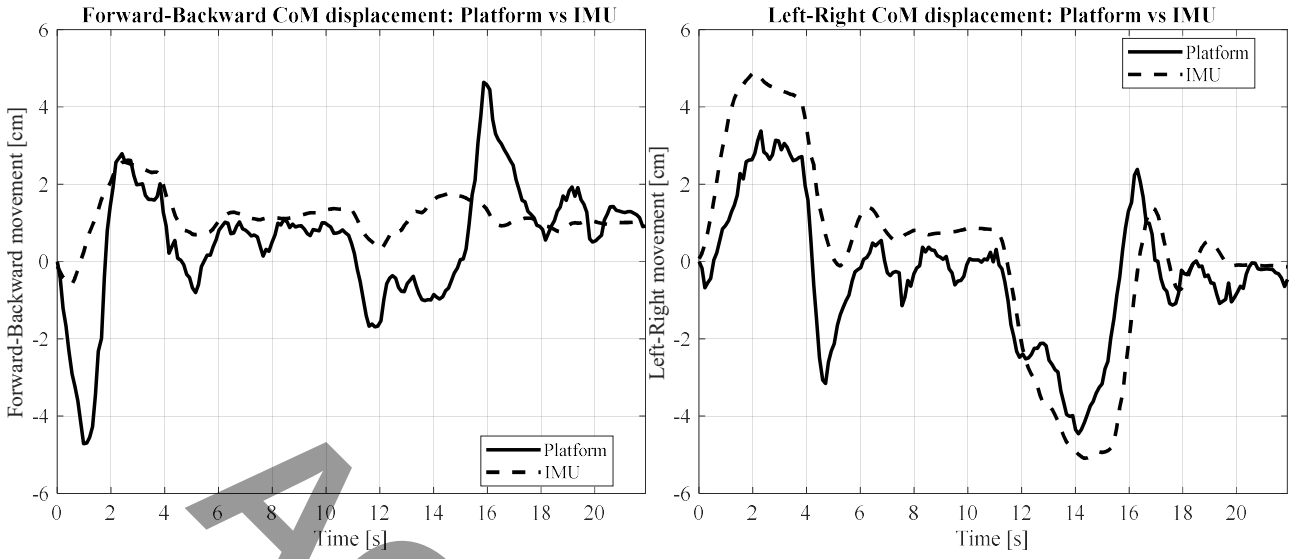


Figure 2. Platform CoP and IMU-derived CoM displacement in sagittal and lateral directions

3.2 Error and Agreement Analysis

Residual error plots indicate that deviations between systems remained relatively stable during the measurement interval, with occasional peaks corresponding to larger movement amplitudes. Residual error plots shown on Figure 3. illustrate the distribution of differences between the systems. No visible systematic drift or progressive error accumulation was observed over time, suggesting temporal stability of the IMU-derived estimates throughout the recording interval. Bland–Altman analysis (Figure 4) showed that most observations were located within the 95% limits of agreement for both movement directions, indicating acceptable agreement between the two systems. The mean bias in the lateral direction was −0.55 cm, with limits of agreement ranging from −3.32 to 2.22 cm. In the sagittal direction, the mean bias was −0.64 cm, with limits of agreement between −3.36 and 2.09 cm. The small negative bias in both directions indicates a slight systematic underestimation of CoM displacement by the IMU system relative to the platform.

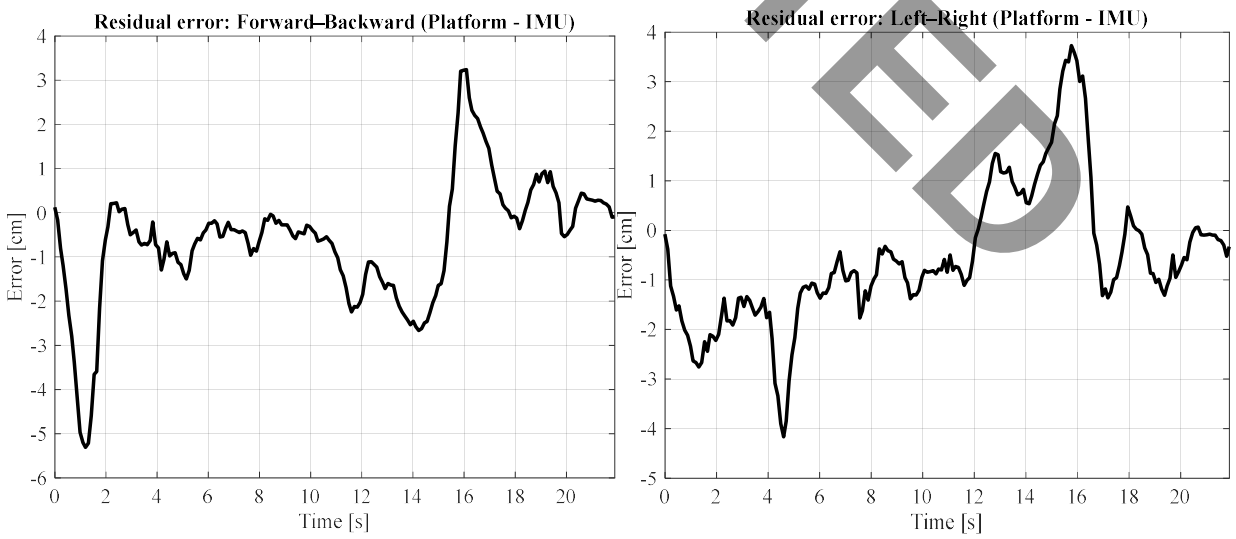


Figure 3. Residual error plots in sagittal and lateral directions

ACCEPTED

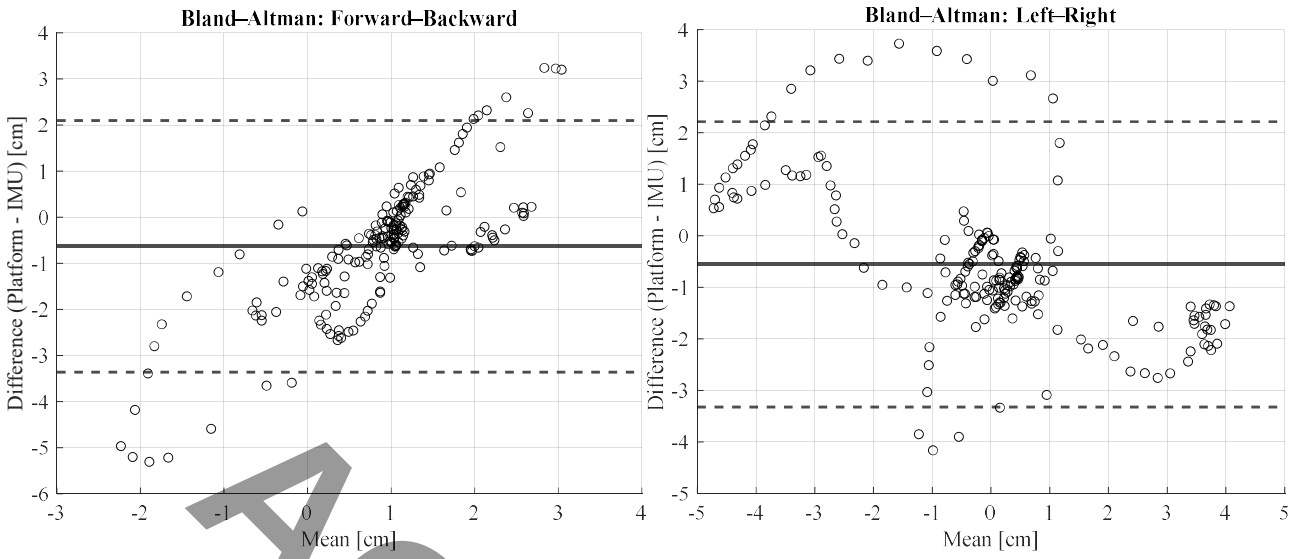
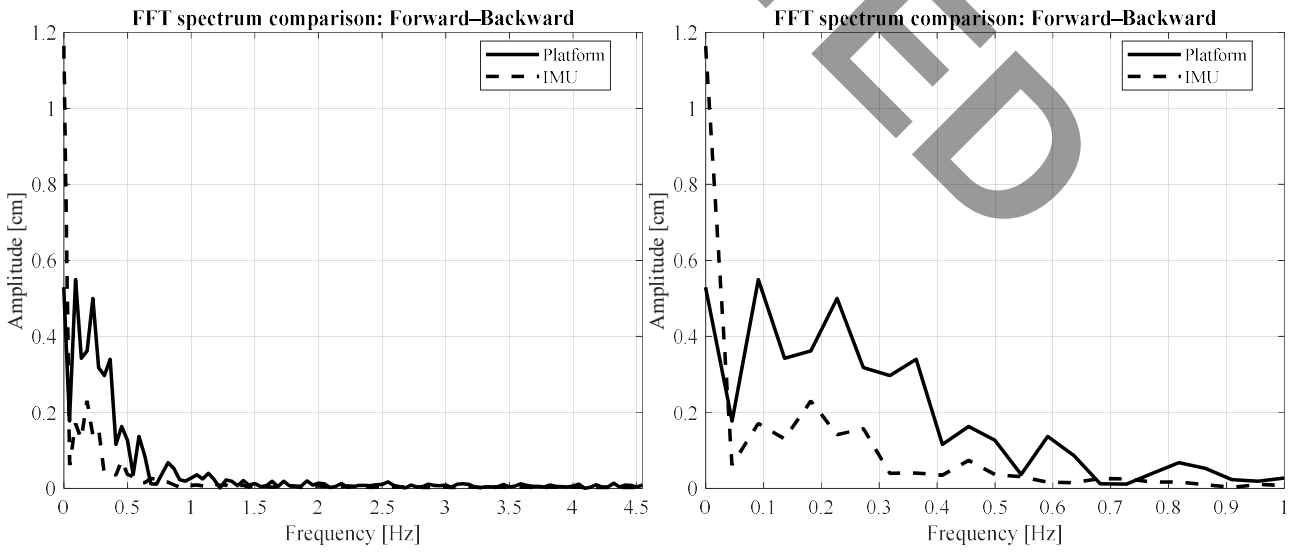


Figure 4. Bland-Altman plots in sagittal and lateral directions

ACCEPTED

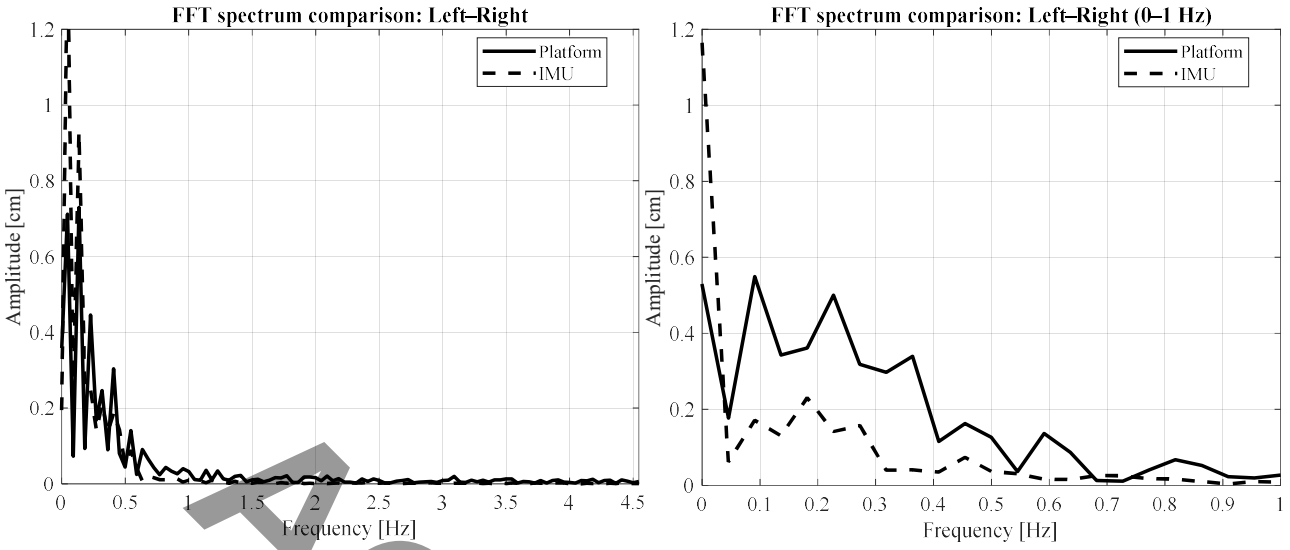
3.3 Frequency-domain comparison

Frequency-domain analysis showed that both measurement systems captured similar dominant low-frequency components of body motion, with most spectral energy concentrated below approximately 0.5 Hz, corresponding to slow, controlled trunk sway during standing rotational movements. The spectra obtained from the IMU system closely followed the spectral distribution of the reference platform, indicating that the reduced torso-mounted sensor configuration preserved the principal temporal dynamics of centre-of-mass motion. Quantitative spectral comparison further supported these observations, as the lateral direction exhibited smaller cumulative spectral deviations (*FFT* difference = 2.82) compared with the sagittal direction (*FFT* difference = 3.63), suggesting slightly stronger agreement between systems for left–right movements than for forward–backward displacement. The spectral difference values represent cumulative absolute deviations across the one-sided frequency spectrum. Frequency spectrum plots in both sagittal and lateral directions are shown in Figures 5 and 6.



a) Entire spectrum *b) Low-frequency region*

Figure 5. Frequency spectrum plot in sagittal direction



a) Entire spectrum *b) Low-frequency region*

Figure 6. Frequency spectrum plot in lateral direction

ACCEPTED

3.4 Integral and Spectral Difference Metrics

The area-under-curve (*AUC*) difference was calculated as a measure of the total absolute deviation between the displacement signals over time. The *AUC* difference represents the cumulative agreement between the two systems across the entire trial duration, rather than instantaneous point-wise error. The computed *AUC* differences indicated comparable overall displacement magnitude differences between the systems in both movement directions (left–right: 27.10 cm·s; forward–backward: 23.91 cm·s), suggesting similar average error magnitudes despite differences in temporal signal correspondence. When normalized to trial duration, these values corresponded to an average absolute deviation of approximately 1.24 cm in the lateral direction and 1.09 cm in the sagittal direction.

4 Discussion

This study evaluated the agreement of centre-of-mass (CoM) displacement during controlled trunk movements between a wearable IMU system and a pedobarographic platform, with the objective of validating a partial IMU configuration for ergonomic assessments in industrial environments. It is important to emphasize that IMU systems and pedobarographic platforms are based on fundamentally different measurement principles. IMU systems estimate body kinematics using segmental accelerations and orientations, from which CoM movement is reconstructed through biomechanical modelling and sensor fusion approaches [16] - [18], whereas pedobarographic platforms quantify pressure distribution beneath the feet and derive the centre of pressure (CoP) as the resultant point of ground reaction forces [6], [13].

The theoretical bridge between these systems lies in the relationship between CoP and CoM. Under quasi-static standing conditions, the CoP approximates the vertical projection of the CoM, as described within inverted pendulum models of postural control [8] - [10]. Gage et al. [8] demonstrated that CoP–CoM dynamics reflect regulation of whole-body sway through modulation of ground reaction forces. However, it is important to note that CoP and CoM are not equivalent quantities. While CoP represents a key parameter important for postural regulation, CoM reflects the mechanical state of mass distribution [9], [10]. Therefore, the agreement observed in this study is focused on biomechanical consistency between biomechanically related but fundamentally different variables.

Most previous validation studies have compared IMU systems with optical motion capture or laboratory-based kinematic reference systems, which are considered a gold standard for motion tracking in laboratory settings. Those comparisons focus mostly on estimating joint angles, segmental orientations and other biomechanical parameters [11], [14], [15]. For example, Desai et al. [29] evaluated IMU-based spatiotemporal gait parameters in comparison with laboratory reference systems and reported strong agreement. While such comparisons assess kinematic reconstruction accuracy, they do not directly address agreement between

kinematic modelling outputs and force-based postural measures. Unlike conventional studies, the present study evaluates IMU-derived CoM displacement in relation to pressure-based CoP measurement, representing a conceptually distinct validation approach. That enables assessment of agreement between biomechanical parameters based on different physical measurement principles.

The results demonstrated strong correspondence between systems in the lateral direction ($r = 0.85$), whereas the sagittal direction showed reduced agreement ($r = 0.49$). Similar directional discrepancies have been reported in trunk-mounted IMU studies. Ferrari et al. [14] demonstrated that agreement between IMU-derived sway parameters and force-platform measurements varies depending on movement direction. Likewise, Noamani et al. [15] reported that trunk-mounted IMUs differ in agreement with force-plate metrics during static balance tasks based on measurement conditions. The lower agreement observed in the sagittal direction in the present study can be explained biomechanically, as sagittal CoM displacement during trunk rotation depends on both trunk motion and compensatory pelvic and lower-extremity contributions, which were not fully represented in the reduced IMU setup. Due to the single-subject design, the numerical agreement values reported in this study should be interpreted as preliminary and specific to the tested subject, movement task, sensor configuration, and data-processing procedure. Therefore, the RMSE, correlation coefficients, AUC differences, spectral differences, and Bland–Altman limits should not be considered population-level validation results. However, the conceptual validation approach, based on comparing IMU-derived CoM displacement with pressure-derived CoP displacement under quasi-static standing conditions, may be transferable to future studies. Similarly, the observed difference between lateral and sagittal agreement provides a plausible methodological indication that partial upper-body IMU configurations are more sensitive to movement directions that depend on unmeasured pelvis and lower-limb contributions.

Despite the reduced sagittal correlation, the RMSE values remained comparable between directions, approximately 1.5 cm, suggesting that overall displacement magnitude error was limited. When expressed relative to total displacement amplitude, the RMSE values corresponded to approximately 19% in the lateral direction and 16% in the sagittal direction. Moreover, the mean absolute deviation between systems remained close to 1 cm in both directions, indicating that the typical instantaneous discrepancy between measurements was moderate in magnitude despite differences in waveform correspondence.

From a practical perspective, the observed RMSE values and mean absolute deviations suggest that the partial IMU setup may be suitable for rapid ergonomic screening, where the objective is to detect larger changes in postural behaviour or identify tasks requiring more detailed assessment. However, these error levels should not be interpreted as sufficient for precise clinical diagnosis, detailed joint-level biomechanical modelling, or detection of small subject-specific changes in sway. Since universally accepted error thresholds for IMU-derived CoM estimation in industrial ergonomic screening are not yet established, the present findings should be interpreted as preliminary evidence of practical feasibility rather than definitive validation of measurement accuracy. Comparable magnitudes of measurement error and configuration sensitivity have been emphasized in IMU validation and methodological studies [11], [18]. For example, Bisi et al. [21] showed that trunk-mounted IMUs can approximate force-platform sway metrics with acceptable levels of error; however, different signal processing methods significantly influenced estimated sway metrics. Van der Kruk and Reijne [11] identified system configuration and application as key factors affecting IMU measurement accuracy. Similarly, Sessa et al. [30] and Beange et al. [31] showed that IMU performance is sensitive to calibration procedures, sensor fusion algorithms, and task characteristics. These findings support the interpretation that partial IMU setups may provide useful information for initial ergonomic screening, although their accuracy depends on sensor configuration, processing strategy, and the movement task being assessed. The systematic bias of approximately 0.6 cm in both sagittal and lateral directions indicates slight underestimation by the partial IMU setup, but its practical relevance should be interpreted cautiously due to the single-subject design and task-specific nature of the experiment.

Frequency-domain analysis further demonstrated that the partial IMU setup reliably captured the dominant low-frequency components of upper-body motion. Quiet standing postural sway, which is a product of balancing through slow, continuous adjustments, is typically characterized by frequency content below approximately 0.5 Hz [32]. The spectral distributions observed in the present study were consistent with this expected range, indicating that the IMU system captured the primary biomechanical features of trunk motion rather than high-frequency noise or artefacts. Foulger et al. [16] pointed out that IMU-derived CoM sway signals closely match reference system characteristics under static stance conditions, while Shepherd and Giblin [23] reported effective use of IMUs in capturing low-frequency movement characteristics in applied

ACCEPTED

performance contexts. Interestingly, the IMU-derived displacement amplitudes were consistently lower than those obtained from the pedobarographic platform. In many IMU validation studies, wearable systems tend to exhibit equal or slightly greater high-frequency components relative to force-platform data due to sensor noise and soft-tissue artefacts [21], [31]. The opposite was observed in this study which can be explained by internal signal-processing within the biomechanical reconstruction software Biomechanics of Bodies, which likely applied a low-pass filter to reduce measurement jitter. Such filtering can attenuate higher-frequency components and reduce overall signal amplitude, which systematically underestimates measurement results, affecting observation in both time and frequency domain analyses.

From an applied perspective, these findings have direct relevance for the deployment of wearable sensing systems in industrial environments. Although full-body motion capture provides higher biomechanical measurement accuracy, the results of this study indicate that a simplified torso-mounted IMU setup was able to reproduce the main characteristics of CoM displacement with moderate magnitude error and minimal systematic deviation.

In industrial contexts where rapid deployment, minimal preparation time, and worker comfort are critical, full-body sensor networks are often impractical due to calibration complexity and operational constraints [19], [20]. Objective ergonomic risk assessment remains essential for improving worker safety and optimizing industrial assembly systems [4], [19], [20]. Simplified configurations have therefore been proposed as scalable alternatives for ergonomic monitoring [21], [22].

An additional implication of the present findings is that, for quasi-static standing conditions, both wearable IMU-based motion capture and pressure-based CoP measurement may serve as practical alternatives to laboratory optical motion capture when field deployment is required. Marker-based systems, although widely considered reference standards, can be difficult to implement in workplace environments due to space limitations, preparation time, and the potential for marker occlusion by equipment, tools, or body segments. In contrast, wearable IMUs do not rely on line-of-sight and can support continuous kinematic estimation during task execution.

Importantly, the present results also suggest that during slow, controlled standing motions, CoP trajectories obtained from a pedobarographic platform can provide a useful proxy for the horizontal projection of CoM, consistent with classical postural control models describing the CoP–CoM relationship under quasi-static conditions. However, this proxy relationship is expected to degrade as movement dynamics increase. It is also important to note that, unlike force platforms that measure full three-dimensional ground reaction forces, pedobarographic platforms measure the vertical pressure distribution, leaving out horizontal shear forces, which limits their ability to fully characterize CoP dynamics during complex movements. During more dynamic tasks involving changes in the base of support, like stepping, gait, or load transfer, CoP is only defined at locations of ground contact and may not capture whole-body mass motion in the same manner, whereas IMU-based CoM estimation remains applicable. With that in mind, selection of sensing modality in industrial ergonomics should be guided by task dynamics and feasibility constraints, with pressure-based systems more appropriate for quasi-static assessments and wearable IMUs more suitable when substantial whole-body motion occurs.

Nevertheless, several limitations must be acknowledged. The study was conducted on a single subject and involved controlled rotational movements under quasi-static standing conditions. As emphasized in CoP–CoM modelling literature [8] - [10], the relationship between these variables becomes more complex under dynamic, multi-planar conditions involving higher accelerations. Therefore, the agreement observed here may not directly generalize to highly dynamic industrial tasks, particularly given that proportional scaling effects may become more pronounced at larger movement amplitudes. Furthermore, trunk-mounted IMU performance depends strongly on sensor placement and processing strategies [16], [18], [30], and future studies should incorporate multi-subject datasets, dynamic task conditions, and comparative analyses across sensor configurations to determine the optimal balance between simplicity, robustness, and biomechanical accuracy in industrial ergonomic applications.

5 Conclusion

This study evaluated the agreement between a partial IMU setup consisting of torso and upper-extremity tracking and a pedobarographic platform for CoM motion assessment. The comparative analysis suggests that a partial IMU setup can provide useful preliminary estimates of CoM displacement under quasi-static

ACCEPTED

conditions, with stronger agreement observed in the lateral direction. The reduced agreement in the sagittal direction is attributed to the absence of pelvic and lower-extremity tracking, which limits the kinematic information available for reconstructing whole-body CoM displacement. The pedobarographic platform, on the other hand, estimates the CoP from the vertical plantar pressure distribution, reflecting whole-body loading patterns.

These findings suggest that a partial IMU setup may provide a practical way to reduce preparation time in industrial ergonomic assessments, particularly when used as a rapid screening tool for quasi-static postural tasks. However, the proposed configuration should not be considered a stand-alone replacement for full-body motion capture or laboratory-grade biomechanical analysis. Considering that industrial settings are characterized by spatial constraints and a need for rapid ergonomic assessments, such simplified configurations may represent a viable screening alternative when full-body MoCap systems are impractical. For stationary or quasi-static working conditions, pedobarographic platforms may also provide useful pressure-based information for postural screening, particularly when full-body kinematic tracking is not required. Within the context of Industry 4.0, partial IMU configurations may offer a viable approach for screening and biomechanical evaluation.

The main limitation of this study is that measurements were conducted on a single subject under controlled quasi-static conditions, which limits the generalizability of the numerical agreement values. Therefore, the reported RMSE, correlation, AUC, spectral, and Bland–Altman results should be interpreted as preliminary and configuration-specific. Future studies should involve larger subject samples, dynamic movement conditions, and different IMU configurations to establish the robustness and broader applicability of simplified IMU-based ergonomic monitoring systems.

ACCEPTED

References

[1] H. Lasi, P. Fettke, H.-G. Kemper, T. Feld, and M. Hoffmann, “Industry 4.0,” *Bus Inf Syst Eng*, vol. 6, no. 4, pp. 239–242, Aug. 2014, doi: 10.1007/s12599-014-0334-4.

[2] J. E. Naranjo, C. A. Mora, D. F. Bustamante Villagómez, M. G. Mancheno Falconi, and M. V. Garcia, “Wearable Sensors in Industrial Ergonomics: Enhancing Safety and Productivity in Industry 4.0,” *Sensors*, vol. 25, no. 5, p. 1526, Feb. 2025, doi: 10.3390/s25051526.

[3] S. Baklouti, A. Chaker, T. Rezgui, A. Sahbani, S. Bennour, and M. A. Laribi, “A Novel IMU-Based System for Work-Related Musculoskeletal Disorders Risk Assessment,” *Sensors*, vol. 24, no. 11, p. 3419, May 2024, doi: 10.3390/s24113419.

[4] S. Digiesi, F. Facchini, G. Mossa, G. Mummolo, “Minimizing and Balancing Ergonomic Risk of Workers of an Assembly Line by Job Rotation: a MINLP Model,” *IJIEM*, vol. 9, no. 3, pp. 129–138, Sep. 2018, doi: 10.24867/IJIEM-2018-3-129.

[5] T. Maillard, “The Three-Dimensional Body Center of Mass at the Workplace under Hypogravity,” *Bioengineering*, vol. 10, no. 10, p. 1221, Oct. 2023, doi: 10.3390/bioengineering10101221.

[6] F. Quijoux et al., “A review of center of pressure (COP) variables to quantify standing balance in elderly people: Algorithms and open-access code,” *Physiol Rep*, vol. 9, no. 22, p. e15067, Nov. 2021, doi: 10.14814/phy2.15067.

[7] Y.-C. Pai and J. Patton, “Center of mass velocity-position predictions for balance control,” *Journal of Biomechanics*, vol. 30, no. 4, pp. 347–354, Apr. 1997, doi: 10.1016/S0021-9290(96)00165-0.

[8] W. H. Gage, D. A. Winter, J. S. Frank, and A. L. Adkin, “Kinematic and kinetic validity of the inverted pendulum model in quiet standing,” *Gait & Posture*, vol. 19, no. 2, pp. 124–132, Apr. 2004, doi: 10.1016/S0966-6362(03)00037-7.

[9] P. Morasso, “Centre of pressure versus centre of mass stabilization strategies: the tightrope balancing case,” R. Soc. *Open Sci*., vol. 7, no. 9, p. 200111, Sep. 2020, doi: 10.1098/rsos.200111.

[10] P. Morasso, A. Cherif, and J. Zenzeri, "Quiet standing: The Single Inverted Pendulum model is not so bad after all," PLoS ONE, vol. 14, no. 3, p. e0213870, Mar. 2019, doi: 10.1371/journal.pone.0213870.

[11] E. Van Der Kruk and M. M. Reijne, "Accuracy of human motion capture systems for sport applications; state-of-the-art review," *European Journal of Sport Science*, vol. 18, no. 6, pp. 806–819, Jul. 2018, doi: 10.1080/17461391.2018.1463397.

[12] L. Chiari, U. D. Croce, A. Leardini, and A. Cappozzo, "Human movement analysis using stereophotogrammetry," *Gait & Posture*, vol. 21, no. 2, pp. 197–211, Feb. 2005, doi: 10.1016/j.gaitpost.2004.04.004.

[13] D. Winter, "Human balance and posture control during standing and walking," *Gait & Posture*, vol. 3, no. 4, pp. 193–214, Dec. 1995, doi: 10.1016/0966-6362(96)82849-9.

[14] L. Ferrari, G. Bochicchio, A. Bottari, A. Scarton, F. Lucertini, and S. Pogliaghi, "Construct Validity of a Wearable Inertial Measurement Unit (IMU) in Measuring Postural Sway and the Effect of Visual Deprivation in Healthy Older Adults," *Biosensors*, vol. 14, no. 11, p. 529, Nov. 2024, doi: 10.3390/bios14110529.

[15] A. Noamani, M. Nazarahari, J. Lewicke, A. H. Vette, and H. Rouhani, "Validity of using wearable inertial sensors for assessing the dynamics of standing balance," *Med. Eng. Phys*., vol. 77, no. 1, pp. 53–59, Mar. 2020, doi: 10.1016/j.medengphy.2019.10.018.

[16] L. H. Foulger, E. R. Reiter, C. Kuo, M. G. Carpenter, and J.-S. Blouin, "Estimating whole-body centre of mass sway during quiet standing with inertial measurement units," PLoS ONE, vol. 20, no. 1, p. e0315851, Jan. 2025, doi: 10.1371/journal.pone.0315851.

[17] M. Nazarahari and H. Rouhani, "Sensor fusion algorithms for orientation tracking via magnetic and inertial measurement units: An experimental comparison survey," *Information Fusion*, vol. 76, pp. 8–23, Dec. 2021, doi: 10.1016/j.inffus.2021.04.009.

[18] J. Li, F. Qiu, L. Gan, and L.-S. Chou, "Concurrent validity of inertial measurement units in range of motion measurements of upper extremity: A systematic review and meta-analysis," *Wearable Technol*, vol. 5, p. e11, 2024, doi: 10.1017/wtc.2024.6.

[19] C. M. Lind, J. A. Diaz-Olivares, K. Lindecrantz, and J. Eklund, "A Wearable Sensor System for Physical Ergonomics Interventions Using Haptic Feedback," *Sensors*, vol. 20, no. 21, p. 6010, Oct. 2020, doi: 10.3390/s20216010.

[20] D. Kelly et al., "Feasibility of Sensor Technology for Balance Assessment in Home Rehabilitation Settings," *Sensors*, vol. 21, no. 13, p. 4438, Jun. 2021, doi: 10.3390/s21134438.

[21] M. C. Bisi and R. Stagni, "IMU-Based balance assessment as an alternative to gold standard COP-based posturography: are we there yet?," *Gait & Posture*, vol. 114, p. S8, Oct. 2024, doi: 10.1016/j.gaitpost.2024.08.023.

[22] S. Lyu, A. Freivalds, D. S. Downs, and S. J. Piazza, "Assessment of postural sway with a pendant-mounted wearable sensor," *Gait & Posture,* vol. 92, pp. 199–205, Feb. 2022, doi: 10.1016/j.gaitpost.2021.11.034.

[23] J. B. Shepherd, G. Giblin, G.-J. Pepping, D. Thiel, and D. Rowlands, "Development and Validation of a Single Wrist Mounted Inertial Sensor for Biomechanical Performance Analysis of an Elite Netball Shot," *IEEE Sens*. Lett., vol. 1, no. 5, pp. 1–4, Oct. 2017, doi: 10.1109/LSENS.2017.2750695.

[24] S. Kongsawasdi, C. Wiboonsuntharangkoon, P. Tajarernmuang, and K. Wantanajittikul, "Feasibility Study of a Prototype Wearable Inertial Measurement Unit for Elderly Postural Sway Assessment," *J Clin Med Res*, vol. 16, no. 4, pp. 174–181, Apr. 2024, doi: 10.14740/jocmr5125.

[25] E. Zemková et al., "Simultaneous measurement of centre of pressure and centre of mass in assessing postural sway in healthcare workers with non-specific back pain: protocol for a

ACCEPTED

cross-sectional study," *BMJ Open*, vol. 11, no. 8, p. e050014, Aug. 2021, doi: 10.1136/bmjopen-2021-050014.

[26] Z. P. Luo, L. J. Berglund, and K. N. An, "Validation of F-Scan pressure sensor system: a technical note," *J Rehabil Res Dev*, vol. 35, no. 2, pp. 186–191, Jun. 1998.

[27] C. Z. Y. Choo, J. Y. Chow, and J. Komar, "Validation of the Perception Neuron system for full-body motion capture," PLoS ONE, vol. 17, no. 1, p. e0262730, Jan. 2022, doi: 10.1371/journal.pone.0262730.

[28] The MathWorks, Inc. MATLAB version R2022b, Natick, Massachusetts, USA, 2022 Available: https://www.mathworks.com

[29] R. Desai, D. Martelli, J. A. Alomar, S. Agrawal, L. Quinn, and L. Bishop, "Validity and reliability of inertial measurement units for gait assessment within a post stroke population," *Topics in Stroke Rehabilitation*, vol. 31, no. 3, pp. 235–243, Apr. 2024, doi: 10.1080/10749357.2023.2240584.

[30] S. Sessa, M. Zecca, Z. Lin, L. Bartolomeo, H. Ishii, and A. Takanishi, "A Methodology for the Performance Evaluation of Inertial Measurement Units," *J Intell Robot Syst*, vol. 71, no. 2, pp. 143–157, Aug. 2013, doi: 10.1007/s10846-012-9772-8.

[31] K. H. E. Beange, A. D. C. Chan, and R. B. Graham, "Evaluation of wearable IMU performance for orientation estimation and motion tracking," in *2018 IEEE International Symposium on Medical Measurements and Applications (MeMeA)*, Rome: IEEE, Jun. 2018, pp. 1–6. doi: 10.1109/MeMeA.2018.8438623.

[32] D. A. Winter, *Biomechanics and Motor Control of Human Movement*, 1st ed. Wiley, 2009. doi: 10.1002/9780470549148.

ACCEPTED