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Evaluating the feasibility of two post-hoc correction techniques for mitigating posture-induced measurement errors associated with wearable motion capture

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ABSTRACT

Wearable motion capture systems are commonly used to measure body kinematics outside of laboratory settings. However, commercially available systems are designed to be used with typically developed adult populations, and assume users begin with a typical standing posture. Individuals with cerebral palsy and other neuromuscular pathologies often present atypical postures that can introduce significant errors in kinematics measurements from wearable motion capture. This study examines two post-hoc correction techniques for rectifying posture-induced errors in kinematic data: (1) Direct three-dimensional realignment of the measured body segment orientations, or (2) adding the initial static joint angle to the dynamic joint angle measurements. Gait kinematics were measured for eight able-bodied participants using a commercial wearable motion capture system. Participants walked with a typical gait, simulated crouch gait, and simulated equinus. The resulting kinematics from the uncorrected and post-hoc corrected trials were compared against simultaneously recorded measurements from an optoelectric motion capture system. Both correction techniques significantly decreased the posture-induced errors in lower-limb joint angle measurements. This work establishes a basis for the application of post-hoc correction techniques, aimed at improving the performance of wearable kinematic measurement systems when used with individuals having non-typical postures.

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1. Introduction

For individuals with cerebral palsy and other movement disorders, three-dimensional assessment of gait kinematics provides clinicians and researchers with valuable information about walking patterns to guide selection and development of rehabilitation treatments [1,2]. While optoelectric motion capture systems are the current gold standard for human movement assessment, wearable motion capture systems are increasingly being used in sports science, ergonomics, and rehabilitation applications. Wearable systems are simpler to use, and are not restricted to a laboratory setting, thus providing clinicians and researchers with greater flexibility to acquire human movement data from everyday activities; such data can guide rehabilitation protocols, design more effective mobility assistive devices, and monitor kinematics during physical activity and exercise [3,4]. Wearable motion capture systems are typically comprised of inertial measurement units (IMUs[1]) which contain tri-axial accelerometers, tri-axial gyroscopes [5], or inertial and magnetic measurement units (IMMUs[2]) having an additional magnetometer to correct for signal drift and errors in the horizontal heading [6]. Whole-body kinematic assessment is achieved when multiple IMMUs (or IMUs) are placed on individual body segments to create a network. Fusion algorithms link the individual sensor signals and define the coordinate transformations between the sensor inertial reference frames and the wearer's anatomic reference frame [7].

This spatial relationship amongst body segments is achieved when the IMMUs are calibrated with the wearer standing in a predefined static posture, so that the body segments (with attached IMMUs) correspond to the specified orientation predefined

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[1] Inertial measurement unit.
[2] Inertial and magnetic measurement unit.

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by the biomechanical model [8,9]. However, individuals with cerebral palsy and similar pathologies are affected by postural deviations such as crouch gait, equinus, and ankle dorsiflexion, among others [10,11] that generate discrepancies between the body segment orientations and the orientations assumed by the biomechanical model, thereby skewing all subsequent joint angle measurements [1,9]. To address this limitation, calibration techniques have been established to achieve a more precise relationship between the sensor and anatomical frames; however these are time-consuming and difficult to apply in practice [6,9,12–15]. An alternatively promising but not very well investigated technique is a post-hoc correction of the kinematic measurements. It was shown in a previous study that a post-hoc correction is able to align the anatomical frame definition of the two systems [16]; therefore a similar procedure could be used to reduce the discrepancies stemming from postural deviations. The techniques capture the static standing posture of the individual to determine the initial joint angles and segment orientations which are applied to the dynamic IMMU-based measurements of gait kinematics as part of a post-hoc analysis.

In this study, we establish and evaluate the accuracy of two post-hoc correction approaches based on a one-time static capture of the true posture of the participant. The first method, termed the orientation correction (OC[3]), directly aligns the IMMU-measured body segment orientations to match the three-dimensional static reference orientations calculated from the one-time static posture capture. The joint angles during gait are then re-estimated using the corrected orientations. The second approach, termed the planar angle correction (PAC[4]), assumes that the joint-angle errors introduced by the posture deviation remain constant throughout gait. The initial joint angles are measured, and then used to correct the IMMU-generated planar joint angles. Therefore, the aims of this study were to 1) to evaluate the performance of the two correction approaches (OC and PAC) in reducing errors induced by postural deviations and 2) determine whether these errors can be approximated and corrected for as constant planar angular offsets or if they have a more complex nature given the three-dimensional misalignment of the body segment. We hypothesized that both approaches (OC and PAC) would significantly reduce posture-induced errors to improve absolute angle measurements, and that the OC method would outperform PAC since it accounts for potential interactions between planes.

2. Methods

2.1. Participants

Eight able-bodied adults with mean ± standard deviation age of 23.8 ± 2.3 years (3 females, 5 males; height, 169.3 ± 9.6 cm; mass, 70.1 ± 13.9 kg) participated. Participants had no physical or cognitive impairments. The study was approved by the institution's research ethics board and all participants provided written consent prior to the study.

2.2. Instrumentation

Each participant was outfitted with the lower-body configuration of the wearable motion capture system MVN BIOMECH AWINDA (Xsens Technologies B.V., Enschede, Netherlands), consisting of seven IMMUs: one on each foot, shank, and thigh, and one on the sacrum, as per manufacturer's guidelines. The Vicon Nexus1.8.5 software connected to seven MX13 cameras (Vicon-Peak, Lake Forest, CA, USA) was used to measure the participants' one-time static posture capture. Sixteen optical markers were placed on the lower body of the participants using the Plug-in Gait configuration [17] (Fig. 1). The Vicon and MVN systems were synchronized for walking trials, and data were sampled at 60 Hz for both systems.

The optoelectric system Vicon Nexus was used as the system independent from the wearable motion capture system, because its kinematics measurements do not require an upright posture calibration, hence it is free from the posture-induced error. Furthermore, as it is considered the industry gold standard for measuring kinematics, it would be an appropriate tool to analyze the post-hoc corrections as a proof of concept.

2.3. Experiment design

Prior to data collection, participants were trained to assume a shallow squat position with their knees bent to at least 20° for crouch gait (CG[5]), and to walk on tip-toes, toe gait (TG[6]). Each participant practiced walking while maintaining the simulated posture until the researcher determined that the posture could be consistently achieved.

Three postures and gait types were used to simulate the deviations: (1) N-Pose (MVN BIOMECH AWINDA system default's calibration posture, a neutral upright posture) and typical able-bodied or normal gait (NG[7]), (2) crouch posture and gait (CG) and (3) tiptoe standing and toe (equine) gait (TG). The static posture was recorded with the optoelectric and wearable systems simultaneously. The wearable system was calibrated using its built-in biomechanical model prior to each walking trial, as a routine step required by the manufacturer. Due to the IMMU's sensitivity to local magnetic fields caused by electronic hardware and steel construction beams in the floor, the participants stood on top of a 50 cm high wooden table for calibration [18]. They assumed a self-selected foot placement, which was marked on the table for repeatability during all subsequent trials. Additionally, the laboratory was mapped [18] and the area with less magnetic distortions was chosen to do the experiments.

Walking trials were performed for the NG, CG and TG conditions. The NG walking condition was to quantify the baseline system agreement between the optoelectric and wearable systems, while the simulated pathological gaits (CG and TG) were used to assess the effects of postural deviations and the efficacy of the two post-hoc corrections. Crouch gait and toe gait were chosen because they primarily exhibit sagittal-plane deviations and are common among many motor pathologies, such as cerebral palsy [10,11]. Gait deviations that would primarily affect the frontal and transverse planes (e.g. genu varum, valgum) were not assessed due to well-documented evidence that even without gait deviations, IMMU-based wearable systems produce significant measurement errors in these secondary joint angles [19–21]. Participants performed three trials in each condition (NG, CG and TG) at a self-selected walking speed.

2.4. Data processing

All data were processed in MATLAB (v. 2016a, The Mathworks, Natick, MA, USA). Each gait cycle was identified using the position data from the right toe and both PSIS markers, using an adapted version of the method described by Zeni et al. [22] that averages the position of the left and right PSIS markers instead of a single-sacrum marker. Despite synchronizing the two systems, data from

[3] Orientation correction.
[4] Planar angle correction.
[5] Crouch gait.
[6] Toe gait.
[7] Normal gait.

Fig. 1. Instrumentation of participant (N-Pose) with IMMUs and markers. Front, back and right-side views. The boxes attached to the legs are foam blocks used to offset the thigh and shank markers, as suggested by the Plug-in Gait model.

each gait cycle had to be temporally aligned by applying a cross-correlation function to the knee-angle profiles to correct for small (< 30 ms) temporal shifts introduced during data collection.

2.5. Body segment orientations and joint angles

The PAC method requires joint angle measurements, which were obtained from both systems for the hips, knees and ankles. Conversely, the OC method requires the orientations of the feet, shanks, thighs and pelvic segments represented in a rotation matrix format. The MVN system provides segment orientations in quaternion format and these were transformed to their rotation matrix equivalents. Additionally, the marker position data from static and walking trials, were exported from the Vicon system for each time point, and lower-body segment orientations were calculated from the filtered marker data using a custom script based on the procedure in the Plug-in Gait manual [17]. The script consists of these following steps:

1. Store the position data as vectors.
2. Estimate the joint centers by using a basic optimization algorithm with the function *isqnonlin* from MATLAB to solve the Chord Function described in the Plug-in-Gait model [17].
3. Define the 3-dimensional components of the body segment based on the position vectors and estimated joint centers.
4. Express the data in 3) in a rotation-matrix format that represents the body segments orientations.

To re-calculate the joint angles from the corrected orientations, the equations from the accepted International Society of Biomechanics joint coordinate system were used [23].

2.6. Correction methods

The OC approach realigns the IMMU-measured segment orientations, $\mathbf{S}$, to match the segment orientations measured by the optoelectric system. The transformation matrices, $\mathbf{T}$ (one per body segment), required to align each IMMU-generated body segment orientation with the optoelectric-system-generated orientations was calculated from the first frame (time $t = 0$; corresponding to the static pose data acquired during the routine calibration), as explained in Section 2.5. The transformation matrices $\mathbf{T}$, account for the difference between the IMMU-generated orientations $\mathbf{S}_0$ (assumed to be in the N-Pose) and the actual posture of the participant during the static calibration. The resulting transformations matrices $\mathbf{T}$, were then applied to all IMMU-measured segment orientations, $\mathbf{S}_f$ of subsequent frames of the walking trial data to obtain the corrected orientations, $\mathbf{C}_f$, using the equation:

$$\mathbf{C}_f = \mathbf{S}_f \mathbf{T}$$

Joint angles were subsequently calculated from each pair of contiguous proximal and distal reoriented body segments, ${}^{prox}\mathbf{C}_f$ and ${}^{dist}\mathbf{C}_f$ respectively, first by using equation:

$${}^{prox-dist}\mathbf{C}_f = {}^{prox}\mathbf{C}^{-1}{}_f\,{}^{dist}\mathbf{C}_f$$

And then parametrizing the joint angles from ${}^{prox-dist}\mathbf{C}_f$, according to the Joint Rotation Convention [23].

For the PAC method, differences in static planar joint angles between both systems during the one-time static posture capture (at time $t = 0$) were determined (offset angles, θ_0) and applied to the dynamic IMMU angles measured during walking (measured angles, M_f), to obtain the corrected angles C_f.

$$\mathrm{C}_f = \mathrm{M}_f + \theta_0$$

2.7. Analysis parameters

Agreement between the optoelectric and wearable joint-angle measurements for each correction and gait type was quantified using a two-step approach. First, kinematics profiles were assessed for shape similarity using the coefficient of multiple correlations (CMC[8]) [24]. Joint angle profiles were each centered and normalized by their maximum joint-angle magnitude prior to the CMC calculations to ensure the resulting values would be independent of any constant mean-offsets between the joint angles measured by the two systems. Besides assessing shape similarity, CMC values were used to evaluate whether the posture-induced error and

[8] Coefficient of multiple correlations.

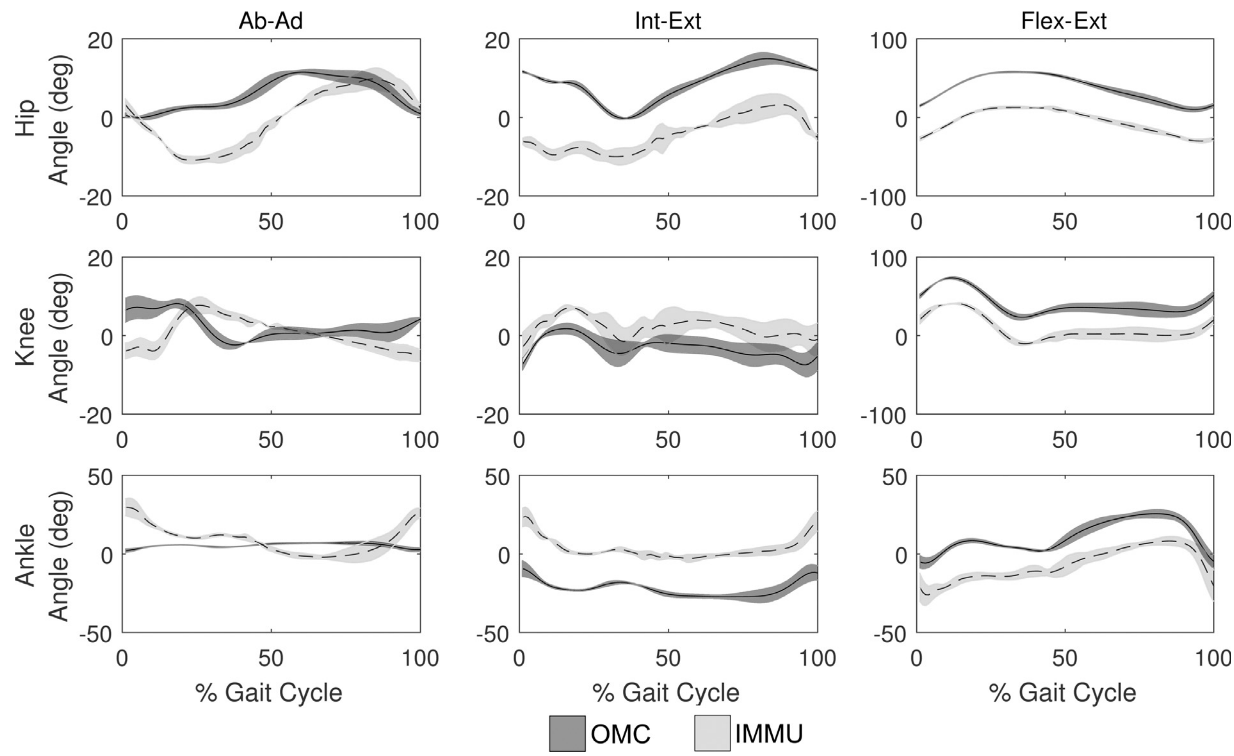

Fig. 2. Typical joint angle profiles measured using both the optoelectric (OMC, solid line) and wearable (IMMU, dashed line) motion capture systems for a single participant walking with crouch gait (CG). The shaded areas represent one standard deviation from the mean. Each row of images corresponds to a single joint, and each column to a joint rotation plane.

the OC have an effect on kinematics waveform. The CMC values of the PAC approach were not calculated, since it does not modify the kinematic waveform.

Secondly, signal offsets were assessed as the mean difference (DIFF[9]) for the gait-cycle-normalized kinematic profiles. The DIFF results, illustrate the mean change in joint-angle measurements. In the normal gait data (NG), DIFF values represent the systemic error, caused by the inherent measurement disagreement between the optoelectric motion capture and IMMU systems. DIFF values of the uncorrected CG and TG data represent the total error caused by both the postural deviations and the systemic error, therefore a reduction in DIFF after applying the corrections can be interpreted as a reduction of the posture-induced error.

2.8. Statistics

A repeated measures analysis of variance (ANOVA) was performed on the CMC and DIFF values for all participants. The major effects of gait type (NG, CG, TG), correction technique (OC, PAC), joint (Ankle, Knee, Hip), and joint rotation (Ab/Adduction, Internal/External, Flexion/Extension) on both metrics (CMC, DIFF) were explored. *P*-values $p < 0.05$ were statistically significant. Tukey HSD post-hoc tests were used determine the significant affecting factors using the Bonferroni correction.

3. Results

An examination of the sample joint angles of a typical gait cycle for both optoelectric motion capture and IMMU measurements (Fig. 2), reveals a high degree of similarity in joint-angle profiles for the Flexion-Extension (Flex-Ext[10]) joint plane, and to a lesser degree for the Abduction-Adduction (Ab-Ad[11]) plane, with the Internal-External (Int-Ext[12]) plane showing mixed results.

CMC values (Fig. 3) were highest for the Flex-Ext angles for all gait types (CMC > 0.90), indicating very good to excellent agreement in the joint-angle profiles measured by the IMMU and optoelectric motion capture systems. The other two planes exhibited significantly lower CMC values ($p < 0.005$), with means of 0.709 ± 0.053 and 0.424 ± 0.012 in the Int-Ext and Ab-Ad planes, respectively. The exception is the Int-Ext angles of the knee joint, which showed good agreement for all corrections and gait types (CMC $> 0.890 \pm 0.065$). The CMC values in the Ab-Ad plane were the lowest overall ($p < 0.013$), with the ankle joint performing especially poorly. There were no differences between right and left leg measurements ($p = 0.337$).

In the Flex-Ext plane, where the CMC values were highest overall, there were no significant differences between the PAC and OC correction techniques for any joint ($p = 0.192$), nor did either correction technique affect the CMC values relative to the uncorrected data ($p > 0.99$). As illustrated by the error-bars in the same figure (Fig. 3), the Ab-Ad and Int-Ext planes evidenced much more variability in the joint angle profiles, but the CMC values remained fairly consistent within each joint-plane combination for both the CG and TG conditions.

Regarding the mean difference, the OC and PAC produced DIFF values in the range of 0 to 10° for all joints in the Flex-Ext plane (Fig. 4). In the majority of cases, both the OC and PAC techniques significantly reduced the mean DIFF ($p < 0.001$) of both simulated gait types relative to the uncorrected measurements. No significant differences were found between correction type ($p = 0.579$). Both corrections also significantly reduced DIFF values for the CG at the knee in the Int-Ext plane ($p < 0.005$), without either outperforming the other ($p = 0.227$). The mean difference of the Flex-Ext and

[9] Mean difference.
[10] Flexion-extension.
[11] Abduction-adduction.
[12] Internal-external rotation.

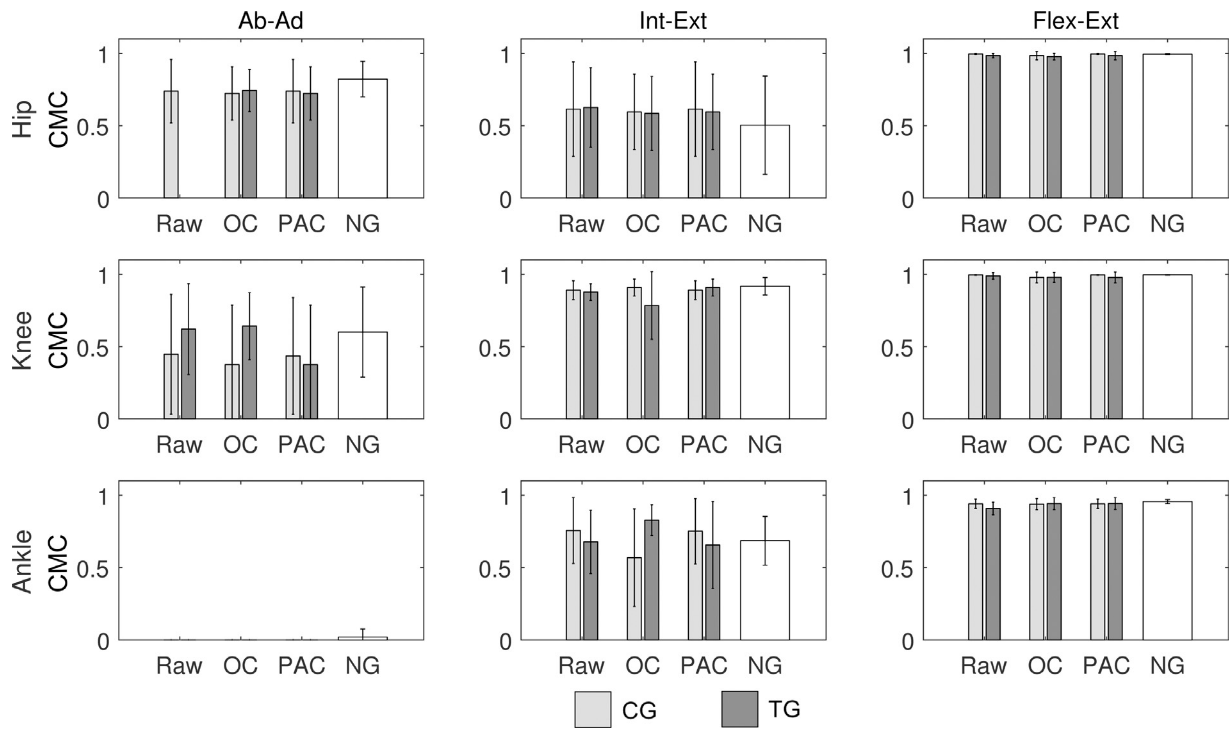

Fig. 3. Coefficients of Multiple Correlation (CMC) values of all participants, based on right side data. Ab-Ad ≡ abduction-adduction; Int-Ext ≡ Internal-External Rotation; Flex-Ext ≡ Flexion-Extension; NG ≡ Normal Gait; CG ≡ Crouch Gait; TG ≡ Toe Gait; OC ≡ Orientation Correction; PAC ≡ Planar Angle Correction; Raw ≡ Uncorrected data.

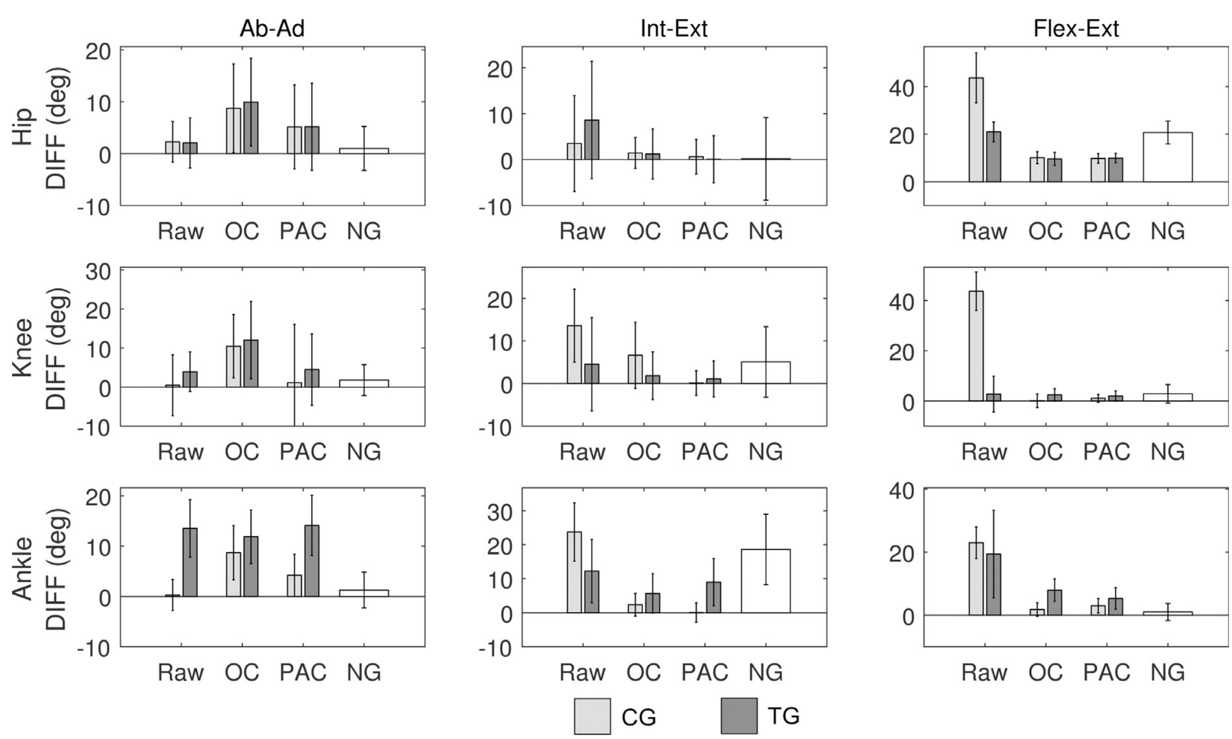

Fig. 4. Difference between optoelectric and wearable motion capture systems (DIFF) values of all participants, based on right side data. Ab-Ad ≡ abduction-adduction; Int-Ext ≡ Internal-External Rotation; Flex-Ext ≡ Flexion-Extension; NG ≡ Normal Gait; CG ≡ Crouch Gait; TG ≡ Toe Gait; OC ≡ Orientation Correction; PAC ≡ Planar Angle Correction; Raw ≡ Uncorrected data.

Int-Ext angles of the knee joint under TG did not change significantly with either correction technique.

Both corrections exhibited significantly lower DIFF values than the NG for the knee and hip of the Flex-Ext plane for CG ($p <$ 0.003), also for the hip and ankle Flex-Ext angles and knee Int-Ext angles during TG ($p < 0.004$). However, the remaining joints and planes showed no significant differences.

4. Discussion

This study examined the kinematic measurement errors of a commercial wearable motion capture system stemming from the potential misrepresentation of the subject's posture in the biomechanical model used for calibration. Using simulated gaits to evaluate postural deviations, these errors were found to be near-constant offsets between the reference and measured joint angles. The comparison of the post-hoc correction approaches, the planar angle correction (PAC) and the orientation correction (OC), revealed the following. In the sagittal (Flex-Ext) plane, both correction techniques similarly reduce postural deviation errors. For gait deviations that occur primarily in the sagittal plane, the simpler PAC technique may adequately achieve the necessary corrections. It is important to note that while in this study the corrections for the PAC technique were based on angles derived from the VICON system, the PAC corrective values can be obtained using much simpler methods, for example from in-plane digital photos during a static pose, by automatically or manually extracting the angles, and applying them as a correction to the planar dynamic wearable measurement IMMU system data. On the other hand, the orientation corrections better capture joint angles in the Int-Ext and Ab-Ad planes, which are more difficult to measure in 2D images.

It is important to note, that the application of the optoelectric system in this study was to provide a gold-standard measurement of posture, to accurately capture potential calibration errors and determine the extent to which these can be corrected. The intent of this study was not to propose that it would be practical to use an optoelectric system, such as a VICON system, for implementation of such corrections. Rather, this study provides important insight as to the origins of the posture-induced errors and demonstrates that substantial improvements in the measurement accuracy could be achieved nearly equally via 2D planar or 3D orientation correction. Since both correction approaches yielded good results in this study, future studies can assess their implementation with more portable systems such as digital photography or goniometer for PAC or technologies similar to the Kinect by Microsoft for PAC and OC methods [25,26].

Comparing the uncorrected data with gait deviations (CG and TG) and able-bodied gait (NG) revealed larger differences in mean joint angles for the uncorrected data for all sagittal plane movements, although a uniform agreement between the joint trajectory profiles remained. These results suggest that the simulated crouch gait and toe gait postural deviations can indeed be considered merely as an offset of measurements in the primary plane of motion (Flex-Ext) and have minimal effects on the joint-angle waveform, suggesting that the nature of the error is not more complex than a shift in the mean joint-angle. Although these results may seem intuitive, prior to this work this evidence was lacking.

The posture-induced error in the Int-Ext and Ab-Ad planes is less evident, due likely to a combination of increased IMMU measurement noise and small ranges of motion in these planes. The CMCs (Fig. 3), and the DIFF values (Fig. 4) show systemic errors and variability in these planes similar to those reported in previous work [19,20]. Hence, the prevalence of these errors supports the decision to limit the evaluation of posture-induced errors to the sagittal plane as was done in this study. However, as IMMU-based measurement technologies are continually improving, to provide more accurate measurement in all planes, future work should aim to evaluate postural deviations in the frontal and transverse planes.

While the study findings provide a strong basis for the use of simple post-hoc kinematic correction techniques, a number of study limitations must be considered. First, while the VICON system is considered the gold standard for determining the errors associated with the wearable system, it should be noted that these systems themselves can vary in accuracy of measurements depending on various factors such as number of cameras, skin motion artifact [27,28], and protocol [29] used among others. Secondly, while it would be desirable to assess postural deviations in all three planes, based on previous studies, and confirmed by the present work, the IMMU based systems have generally demonstrated poor measurement performance in the frontal and transverse planes [19–21].

The use of simulated gaits is potentially another limitation, since these represent approximations of clinical manifestations. However, the use of simulated postures can provide a first-approximation and proof-of-concept for the proposed corrections that is non-patient specific and therefore more generalizable. Other studies have evaluated the gait of able-bodied adults simulating pathological gait, and have found that these simulated pathologies provide an acceptable first-approach scenario for studying the effects of non-typical postures on gait kinematics [30,31]; future studies should expand on the implementation of the corrections on pathological data.

In this study the direct comparison of joint-angle profiles for normal gait between the optoelectric and wearable systems showed good measurement agreement in the sagittal plane based on the CMC values ($CMC > 0.90$) and angle measurement errors less than 5°. The lone exception is the hip joint, which exhibited mean errors of $20.7 \pm 4.8°$. However, this discrepancy can be attributed to the different definitions of the pelvis orientation employed by the two systems [21], and despite the initial angle difference, the net hip angle measurements remained nearly identical for the two systems throughout the gait cycle ($CMC > c0.98$). A previous study [16] showed that the corrective techniques not only compensated for postural deviations, but also inherent model differences in the two systems. However, as in the present study the purpose was to assess the postural deviation alone, the anatomical frame error was considered as part of the systemic error, which was calculated from the normal gait trials as a reference. In practice, implementing the method proposed by Li and Zhang [16] as a complement to either of the two post-hoc corrections in this work, could further improve kinematic measurements, particularly in the Ab-Ad and Int-Ext planes. However, the analysis of these two planes needs to be addressed in a separate study.

In conclusion, the findings suggest that the measurement errors in the case of non-typical postures stemming from misrepresented models during calibration can be approximated as an offset and that these posture-induced errors can be significantly decreased using either the OC or the potentially simpler-to-implement PAC methods. This study represents a basis for development of an effective, efficient and convenient alternative method for dealing with postural deviations to achieve higher accuracy when measuring kinematics using IMMU-based systems. Future work should focus on the use of digital photography or goniometers for obtaining the initial joint angles required for the PAC in case gait deviations and measurements of interest are primarily in the sagittal plane. Similarly, technologies such as Kinect by Microsoft may useful for obtaining initial joint angles in three-dimensions, and should be explored as low-cost and readily available tools for implementing the OC method with individuals exhibiting multi-plane postural deviations. Continuation of this research has particular importance in the assessment and rehabilitation of individuals with both acute and chronic neuromusculoskeletal-related movement disorders.

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Ethical approval

The Holland Bloorview Research Ethics Board as well as the Office of Research Ethics of the University of Toronto, approved the research presented in this paper under the file numbers 14–512 and 31,504, respectively.

Conflict of interest

The authors have no conflicts of interest to disclose.

Supplementary material

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.medengphy.2019.06.013.

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