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Assessment of the measurement accuracy of inertial sensors during different tasks of daily living

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ABSTRACT

The low cost and ease of use of inertial measurement units (IMUs) make them an attractive option for motion analysis tasks that cannot be easily measured in a laboratory. To date, only a limited amount of research has been conducted comparing commercial IMU systems to optoelectronic systems, the gold standard, for everyday tasks like stair climbing and inclined walking. In this paper, the 3D joint angles of the lower limbs are determined using both an IMU system and an optoelectronic system for twelve participants during stair ascent and descent, and inclined, declined and level walking. Three different datasets based on different hardware and anatomical models were collected for the same movement in an effort to determine the cause and quantify the errors involved with the analysis. Firstly, to calculate software errors, two different anatomical models were compared for one hardware system. Secondly, to calculate hardware errors, results were compared between two different measurement systems using the same anatomical model. Finally, the overall error between both systems with their native anatomical models was calculated. Statistical analysis was performed using statistical parametric mapping. When both systems were evaluated based on the same anatomical model, the number of trials with significant differences decreased markedly. Thus, the differences in joint angle measurement can mainly be attributed to the variability in the anatomical models used for calculations and not to the IMU hardware.

1. Introduction

Stair navigation and inclined walking have been analysed using a small number of steps or a treadmill (Kimel-Naor et al., 2017; Bergmann et al., 2009). Unfortunately, laboratory conditions do not adequately represent real-world situations (Kimel-Naor et al., 2017). The increasing availability of inertial sensors (IMUs) for motion analysis enables measuring complex tasks in real-world environments. Inertial systems excel due to their low cost, ease of usability and unrestricted measurement volume (Tao et al., 2012). Unfortunately, these advantages are currently jeopardised by measurement discrepancies between wearable and optoelectronic systems.

These discrepancies have been addressed in different studies using commercial IMU systems for level walking (Mundt et al., 2017; Nüesch et al., 2017; Zhang et al., 2013; Ferrari et al., 2010; Cloete and Scheffer, 2008; Picerno et al., 2008), long-term ergonomic tasks (Robert-Lachaine et al., 2016; Kim and Nussbaum, 2013), stair climbing (Mundt et al., 2017; Zhang et al., 2013; Bergmann et al., 2009) and inclined walking (Mundt et al., 2017). All aforementioned studies have shown a good agreement in angle waveform but an offset in the sagittal plane. In the non-sagittal planes, the angle waveforms did not agree as well. Possible explanations are different anatomical models used by the systems or limitations of the IMUs themselves. Anatomical models define anatomical rotation axes (Cappozzo, 1984) and thereby influence the results of the joint angle calculation. The application of the same anatomical model to optoelectronic and IMU data significantly improved the conformity in angle waveform and eliminated the offset in the sagittal plane (Ferrari et al., 2010; Kim and Nussbaum, 2013; Robert-Lachaine et al., 2016).

With this study, we aim to provide further insight into the comparability of IMU and optoelectronic systems. The lower limb joint angle waveforms measured by both systems during level walking, inclined walking and stair climbing are compared. The error is further separated into the absolute error, the error based on the anatomical model and the error based on the IMUs as such. Findings of previous studies are expanded to more complex motion tasks that have not been analysed in detail before. This study hypothesises that (1) the accuracy of the joint angle waveforms for inclined walking and stair climbing is higher than for level

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walking due to the higher ranges of motion that can be captured more efficiently by the IMUs than small motions and (2) the anatomical model accounts for the majority of deviations in joint angle waveforms.

2. Methods

2.1. Experimental set-up

Twelve healthy subjects (5 female, 26.9 ± 2.3 years, 70.8 ± 13.2 kg, 171.9 ± 10.2 cm) participated in this study that was approved by the Ethical Committee of the German Sport University Cologne. All participants provided their informed written consent. Each subject performed ten walking trials of the following nine conditions in the laboratory: straight level walking at five different speeds (0.8 m s^{-1}, 1.1 m s^{-1}, 1.4 m s^{-1}, 1.7 m s^{-1}, 2.0 m s^{-1}) on a 5 m walkway; stair ascend and descent (5 steps, 0.2 m height, 0.3 m depth, 0.74 m width); inclined and declined walking on a slope of 20% (Karamanidis and Arampatzis, 2009; Komnik et al., 2016) at a self-selected speed. The motion was captured simultaneously by ten infrared cameras (100 Hz, VICON™, MX F40, Oxford, UK) and an IMU system (100 Hz, MyoMotion, Noraxon U.S.A. Inc., Scottsdale, Arizona, USA, see Appendix A, Harrington et al. (2007)) (cf. Fig. 1).

2.2. Anatomical models

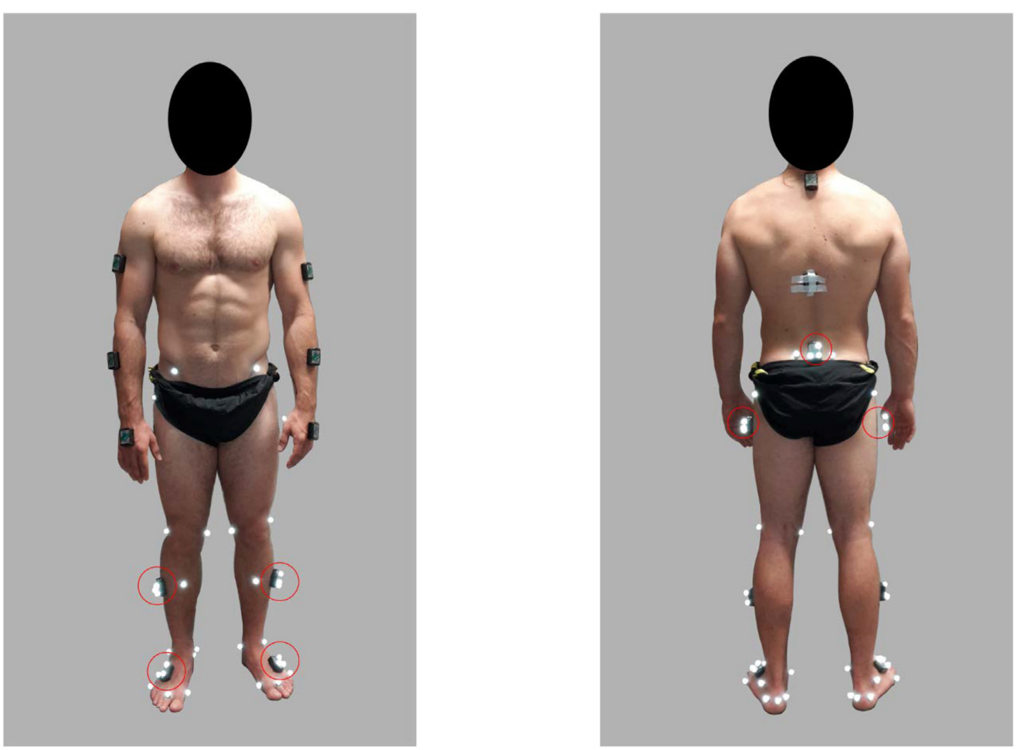
The optoelectronic system used an anatomical model based on the recommendations of the International Society of Biomechanics (ISB) (Wu et al., 2002). Therefore, 28 reflecting markers were attached to defined bony landmarks to create a seven segment rigid model. The segment and rotation coordinate systems were defined according to Wu et al. (2002) and Pennock and Clark (1990). The detailed marker positions are displayed in Table 1.

The built-in model of the IMU system was based on the sensor orientations determined from the IMUs using a Kalman filter. These orientations were given in the global reference system of the IMU system (MyoMotion, Noraxon U.S.A. Inc., Scottsdale, Arizona, USA), based on the direction of the magnetic north (x-axis) and the direction of the gravity (z-axis). The y-axis was defined as perpendicular to x and z, and pointing to the east. The calibration, i.e. the sensor-to-segment alignment, was performed with the subject standing in a neutral position with straight legs. It was assumed that each segment and joint is aligned during this reference pose. Therefore, the offset rotation from the sensor in the global frame to the target, a prior known segment pose, was determined. Based on this offset rotation the segment and rotation coordinate systems were defined (Oberländer, 2015). To determine possible causes of error, the joint angles of the IMU system were additionally calculated using the ISB anatomical model.

The calculation of joint angles was based on the segment and rotation coordinate systems defined by the anatomical models (Robertson et al., 2013).

2.3. Data analysis

To synchronise both measurement systems, a transistor-transistor logic signal was exported. Marker trajectories were filtered using a zero-lag fourth-order low-pass Butterworth filter with a cut-off frequency of 6 Hz (Robertson et al., 2013). All angles were time normalised to 0–100% gait cycle, lasting from one initial contact to the next. The initial contact of each step was determined by the foot contact algorithm (Maiwald et al., 2009). For each par-



(a) Frontal view (b) Dorsal view

Fig. 1. 49 reflective markers were used in this setup: 28 markers were attached to the participants' body according to the ISB anatomical model, additional 21 markers were attached to the IMUs to provide a reference of the IMUs orientation in the global reference frame. These markers were not used for further analysis. The IMU system consists of 15 IMUs. Only the information of the sensors attached to the lower body (circled) were used for further analysis.

Table 1
Position of the 28 reflective markers attached to the body to determine the segments orientations and rotation axes according to the ISB recommendations.

RASI, LASI	Anterior superior iliac spine
RPSI, LPSI	Posterior superior iliac spine
RTRO, LTRO	Lateral prominence of the greater trochanter external surface
RKNE, LKNE	Lateral femoral epicondyle
RKNEM, LKNEM	Medial femoral epicondyle
RTIB, LTIB	Anterior border of the tibial tuberosity
RANK, LANK	Distal apex of the lateral malleolus
RANKM, LANKM	Distal apex of the medial malleolus
RCAL, LCAL	Lateral calcaneus
RCALM, LCALM	Medial calcaneus
RHEEL, LHEEL	Upper ridge of the calcaneus posterior surface
RMT1, LMT1	Dorsal aspect of first metatarsal head
RMT5/LMT5	Dorsal aspect of fifth metatarsal head
RTOE/LTOE	Dorsal aspect of second metatarsal head

ticipant, the first step of each trial, which leads to 10 steps for each condition, were used for further analysis. Participants with less than 10 steps for one condition were excluded. This led to the exclusion of one participant for the gait velocities of 1.4 m s^{-1} and 1.7 m s^{-1} and two participants for 2.0 m s^{-1}. This led to a total number of 120 steps for level walking at 0.8 m s^{-1} and 1.1 m s^{-1}, stair negotiation and inclined walking, 110 steps for level walking at 1.4 m s^{-1} and 1.7 m s^{-1} and 100 steps for level walking at 2.0 m s^{-1}.

Based on the marker trajectories and the IMU orientations, three sets of joint angles were assessed using MATLAB (Version 2018a, The MathWorks, Inc., Natick, Massachusetts, USA):

S1 Optoelectronic angles using the ISB anatomical model.
S2 IMU angles using the IMU anatomical model.
S3 IMU angles using the ISB anatomical model.

A detailed mathematical description on the coordinate system transformations used to calculate the joint angles of the IMU system using the ISB anatomical model (S3) is given in the Appendix.

For later comparison tests T1 and T2 were defined to assess different possible errors of the IMU system compared to the optoelectronic system. T1 is used to assess the software error, while T2 displays the total error:

T1 different measurement systems and the same anatomical model (S1 vs. S3).
T2 different measurement systems and different anatomical models (S1 vs. S2).

2.4. Statistical analysis

The repeatability of the three datasets was assessed using the Intraclass Correlation Coefficient (ICC) with 95% confidence intervals (Nüesch et al., 2018; Robert-Lachaine et al., 2017) and the Standard Error of Measurement (SEM) (Robert-Lachaine et al., 2017). A linear mixed model with subject as random factor was used to determine whether there are significant subject effects. To analyse whether the gait velocity influences the comparability of the two systems, a one-way repeated measures ANOVA was performed on the mean root-mean-square error normalised to the range of the data (nRMSE) comprising all subjects at each of the five gait velocities.

One mean nRMSE value was calculated to summarise the differences in angle waveforms of two corresponding steps. Thereby, single steps could be analysed. SPM analysis was performed to detect statistical differences in angle waveforms. Due to non-normal distributed data and inhomogeneous variances, the SPM non-parametric one-way repeated measures ANOVA ($\alpha = 0.05$) was used. ANOVA was conducted for each condition, movement and joint, combining all three datasets. This resulted in an input matrix of [(datasets, subjects, trials) $\times$ 101 time frames]. In case of statistical differences, additional ANOVA analyses were conducted as post hoc test using datasets S1 and S3 for T1 and S1 and S2 for T2 to account for differences in variance of the datasets (SPM, v.M0.4, www.spm1d.org (Nichols and Holmes, 2001)). The α-level was adjusted using Bonferroni correction. To summarise the results of the SPM analysis, the accumulated total significant differences for each test were determined. Since each curve is normalised to 100% of the gait cycle, this value shows in how many percent of the gait cycle statistically significant differences occur (cf. Fig. 2).

3. Results

The one-way repeated measures ANOVA did not show significant differences (T1: p = 0.242, T2: p = 0.057) between the level walking trials. Therefore, all 560 trials were summarised and referred to as the parameter "gait". The analysis of the linear mixed model revealed significant subject effects for all motion tasks ($\alpha < 0.05$).

3.1. Repeatability

The repeatability of dataset S3 was lower than the repeatability of the other two datasets (S1: ICC = 0.974 ± 0.010, SEM = 1.43 ± 0.42°, S2: ICC = 0.930 ± 0.026, SEM = 2.28 ± 0.65°, S3: ICC = 0.810 ± 0.068, SEM = 6.24 ± 1.94°).

3.2. Normalised root-mean-squared error

The nRMSE showed comparable results for all five movement conditions (T1: level walking 13.2 ± 7.5%, inclined walking 11.7 ± 5.6%, declined walking 9.1 ± 4.9%, stair ascending 11.9 ± 6.3%, stair descending 9.9 ± 4.4%; T2: level walking 29.3 ± 18.0%, inclined walking 21.7 ± 12.9%, declined walking 23.2 ± 14.8%, stair ascending 21.9 ± 11.0%, stair descending 25.2 ± 11.4%). The overall nRMSE was smaller for T1 than for T2 (T1: 11.3 ± 2.3%, T2: 24.3 ± 5.3%), showing smaller deviations between angle waveforms based on the same anatomical model. The nRMSE was smallest in the sagittal plane for both tests (T1: 5.8 ± 1.7%, T2: 16.9 ± 5.1%), while it was larger in the transverse (T1: 10.6 ± 2.5%, T2: 25.0 ± 3.2%) and frontal plane (T1: 17.0 ± 2.7%, T2: 30.9 ± 7.6%). The distribution of the nRMSE for each subject is displayed exemplarily for stair ascending in Fig. 3. For all conditions, the nRMSE was smaller in the sagittal and transverse plane when using the same anatomical model, but this effect was not found consistently in the frontal plane. Additionally, some subjects had distinctly larger standard deviations than others.

3.3. SPM analysis

The ANOVA analyses showed statistically significant differences between the three datasets S1, S2 and S3. T1 – the difference between S1 and S3 – corresponds to the hardware error only, while T2 – the difference between S1 and S2 – describes the overall error. Thus, the difference between these two tests is a measure of the software error (cf. Figs. 2 and 4). The accumulated significant differences for all conditions was 24.5% for T1 and 64.4% for T2. For the different conditions, the maximum accumulated significant difference was 36.1% (level walking) for T1 and 68.4% (declined

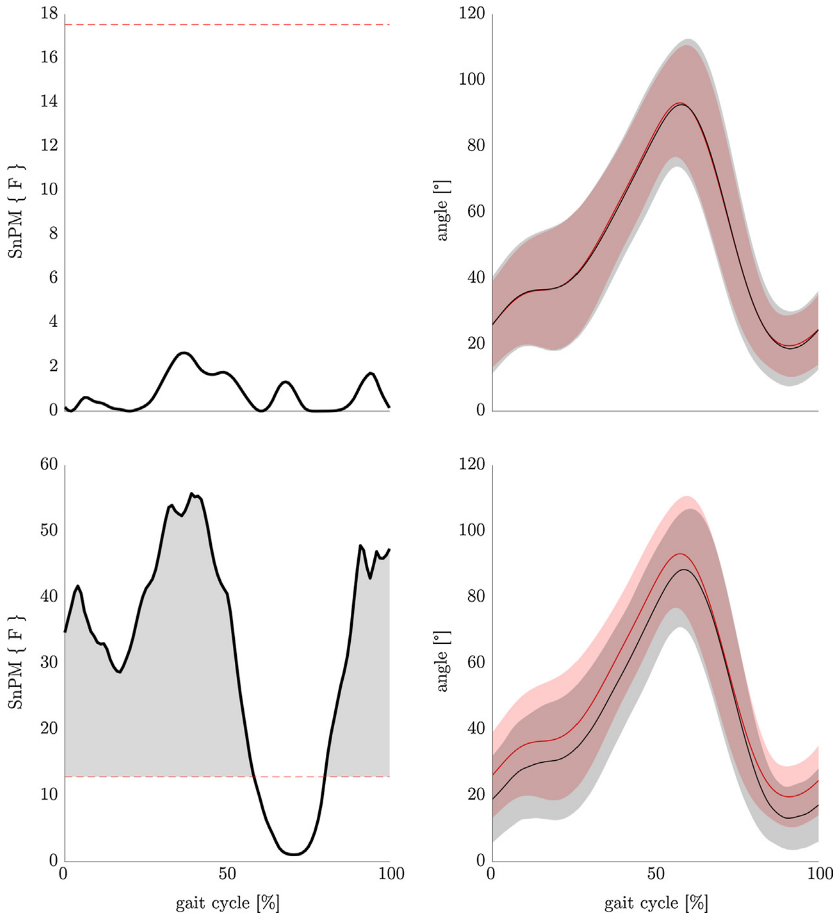


Fig. 2. SPM results for knee flexion during stair descending for T1 (top left) and T2 (bottom left) and the corresponding mean and standard deviation of the joint angles (right). The grey shaded area in the SPM plots indicates significant differences in the angle waveforms. The accumulated significant difference is the percentage of the gait cycle where the F-curve exceeds the threshold (red dashed line). In case of T2 the accumulated difference is a total of 79%. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

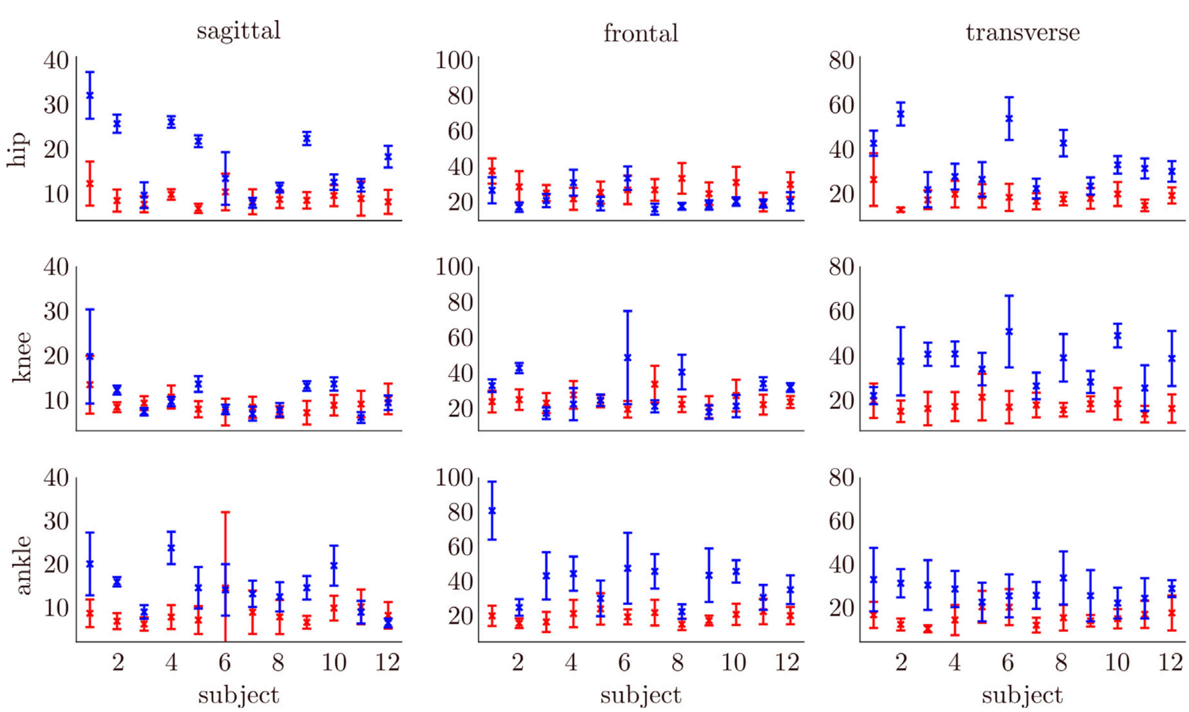


Fig. 3. Mean and standard deviation of the normalised RMSE for each subject for stair ascending. T1 is displayed in red, while T2 is displayed in blue. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

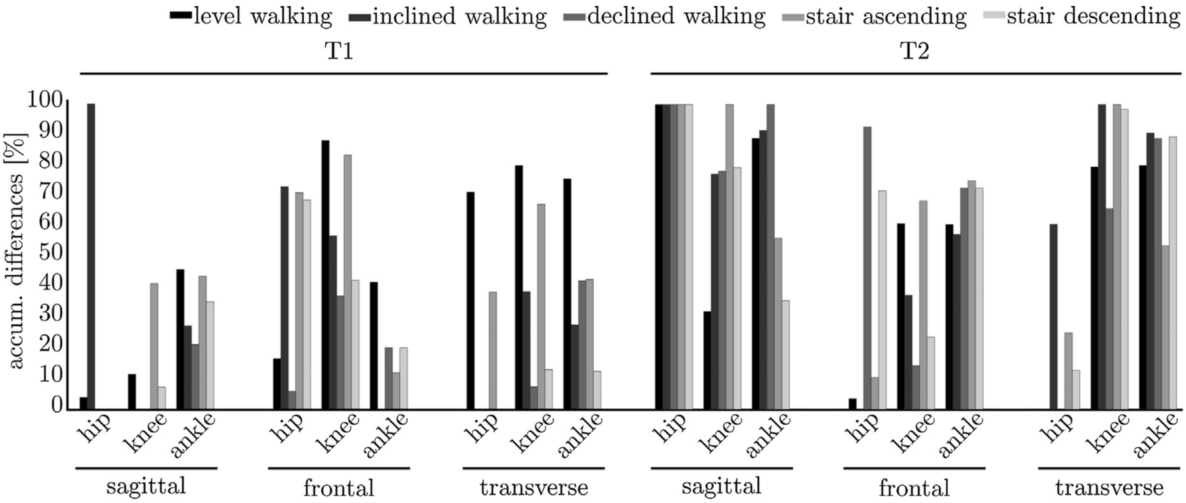


Fig. 4. Accumulated total significant difference for the different joints, motion planes and movement tasks for T1 and T2. The total is defined as the accumulated percentage of waveforms showing significant differences. The reduction of the total comparing the two tests illustrates the software error, i.e. the error due to the anatomical model used. The bars of T1 illustrate the hardware error, while the bars of T2 display the sum of hardware and software error.

walking) for T2. The smallest accumulated significant difference was 15.1% (stair descending) for T1 and 56.2% (level walking) for T2. The smallest number of significant differences was found in stair ascending, the highest in declined walking. The improvement of comparability – a measure for the influence of the software on curve similarity – was only 20.2% for level walking. This effect was much stronger for the other conditions: for declined walking, the accumulated significant difference was decreased by 41.5%, for inclined walking by 57.0%, for stair ascending by 32.3% and for stair descending by 48.7% (cf. Fig. 4).

4. Discussion

This study assessed the joint angle measurement of an IMU system compared to an optoelectronic system and analysed the influence of the anatomical model and of the technology itself on the joint angle waveforms for different activities of daily living. The results supported hypothesis (2): deviations in joint angle waveforms are mainly based on differences in the anatomical model (cf. Fig. 4). This affected mainly inclined walking and stair ascending and, to a limited magnitude, level walking. The IMU system with its built-in software was capable of determining joint angles during level walking more accurately than for the other tasks. Hypothesis (1) was disproved: the accuracy for level walking was higher than for the other motion tasks. The adaptation of the anatomical model showed a greater improvement in comparability for all tasks than level walking, which may imply that the IMU system is optimised for level walking. These results were supported by the nRMSE: for T1 the nRMSE was lower for all tasks compared to T2.

Dataset S3 was calculated based on a constant offset matrix, which might explain the lower repeatability of this dataset compared to the other ones because it neglects soft tissue movements. While the SPM analysis considered all trials of one subject, the nRMSE was calculated for single trials and evaluated for each subject individually. This analysis depicted that there are some subjects showing a large standard deviation in the nRMSE, especially for T2. This higher variance can be caused by the calibration procedure of the IMU system that is dependent on a fixed pose. This pose might be reproduced more or less reliably by different subjects. Due to the high variance in the nRMSE data, subject effects may be generated.

Nevertheless, the error values of the presented study regarding level walking correspond to the results of Nüesch et al. (2017) and are even lower than reported by Cloete and Scheffer (2008). Only Robert-Lachaine et al. (2016) reported lower values analysing ergonomic tasks. This might be attributed to the different IMU systems used in these studies, which have a higher repeatability (Robert-Lachaine et al., 2017).

In level walking, the main difference in joint angles in the sagittal plane was the offset between the angle waveforms and not the progression itself (Mundt et al., 2017; Nüesch et al., 2017; Robert-Lachaine et al., 2016; Ferrari et al., 2010; Cloete and Scheffer, 2008). This study extended these findings to the more complex motion tasks, i.e. inclined and declined walking and stair ascending and descending. Zhang et al. (2013) found comparable differences during stair ascending and descending to those of level walking in all three motion planes. They stated that adduction-abduction and internal-external rotation cannot be compared to the angles determined by an optoelectronic system easily because of the large differences in motion planes with a small range of motion. Unfortunately, they did not present RMSE values that can be compared to the results of this study. Bergmann et al. (2009) noted a smaller RMSE for stair climbing than in the presented study for the sagittal plane motion, but they placed the optoelectronic markers on the IMU, whereby IMU and markers experience the same soft tissue movement, while we used anatomical landmarks to create a better reference to the optoelectronic system. Both aforementioned studies did not investigate the influence of the anatomical model. Due to the interdependency of the joint angles in all three motion planes, a coordinate transformation is necessary to adequately remove the offset visible in the sagittal plane and to improve the results in the other two motion planes. The alignment of the segment and joint coordinate systems is even more important for the more complex tasks analysed in this study. These motions are characterised by higher ranges of motion than level walking. Since the accuracy of the measurement technology as such is more suitable for larger ranges of motion, the coordinate system alignment is even more important. These results emphasise the need of better conformity and documentation of anatomical models used for motion analysis systems to improve the comparability between different systems. Concurrent software updates of IMU-based systems show the effort of the manufacturers to overcome these problems.

5. Conclusion

The presented study showed that the main differences in joint angle measurements can be attributed to the anatomical models used by different measurement systems. The presented method improves the comparability of the IMU system to an optoelectronic system and the application of IMUs in the analysis of more complex motion tasks can be extended. Particularly due to the high amount of effort that is required for post-processing optoelectronic data from stair ascending and descending, the utilisation of IMUs is of high interest. The lower costs, easier use and unrestricted measurement volume and location will make IMU systems a valuable tool, especially for clinical applications. However, the necessity of calibration postures, the limited comparability to optoelectronic system and the lower precision of the measurement systems are still a drawback. Research on how to further improve the comparability and simplify the usability of IMU systems for motion analysis is urgently needed.

6. Limitations

In this study, 12 subjects without any movement disorder were analysed. The small number of participants and their homogeneity is a limitation. The IMU system used in this study showed a lower repeatability than those systems used in other studies. These measurement deviations may originate from the fixation, size and weight of the sensors. The sensors are exposed to soft tissue movements that cause wobbling of the sensor and affect the internal IMU. This might lead to the overestimation of the joint angles in the non-sagittal planes. It would be advantageous to apply the method from this study to another measurement system for further verification. Additional drawbacks of the proposed method are the use of a constant offset matrix and the necessity of a second optoelectronic system. As soon as the position of the IMU changes, either due to soft tissue movements or fixation problems, the offset matrix becomes invalid and the alignment erroneous. There are frequent software releases by Noraxon that aim to overcome this problem. Another drawback of the IMU system is the calibration approach: firstly, it is dependent on the subject to achieve the calibration posture and, secondly, coordinate systems are defined on the segment orientation only. Thereby, the IMU system disregards possible physiological reasons for deviations, e.g. varus or valgus positions of joints. Additionally, the sensors might be influenced by magnetic distortion, which will lead to an incorrect offset rotation. Both limitations strongly influence the joint angle calculation (Oberländer, 2015).

Conflict of interest statement

The authors have no conflicts of interest.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.jbiomech.2018.12.023.

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