THE INFLUENCE OF MOTION TASKS ON THE ACCURACY OF KINEMATIC MOTION PATTERNS OF AN IMU-BASED MEASUREMENT SYSTEM

Marion Mundt[1], Anna Wisser[2], Sina David[2], Thomas Dupré[2], Valentin Quack[3], Franz Bamer[1], Markus Tingart[3], Wolfgang Potthast[2,4], Bernd Markert[1]

Institute of General Mechanics, RWTH Aachen University, Aachen, Germany [1]
Institute of Biomechanics and Orthopaedics, German Sport University Cologne, Cologne, Germany [2]
Department of Orthopaedic Surgery, Uniklinik RWTH Aachen, Aachen, Germany [3]
ARCUS Clinics Pforzheim, Pforzheim, Germany[4]

The amount of inertial measurement unit (IMU)-based motion analysis systems on the market rises. Due to lower costs and easier usage, these systems offer an alternative to optoelectronic systems. Previous studies showed differences between IMU-based and optoelectronic systems during gait. This study evaluates the measurement accuracy of 3D lower limb joint angles of the IMU-based measurement system MyoMotion during different motion tasks. Therefore, 12 healthy subjects are investigated during level walking, sloped walking and climbing stairs. Differences in the results are observed in the biomechanical model. Moreover, the movement velocity causes differences between both measurement systems. Especially during more complex and faster movements, these differences increase.

KEY WORDS: IMU, gait analysis, MyoMotion, accuracy.

INTRODUCTION: Inertial measurement units (IMUs) aim to offer a low-cost and easy-to-use alternative to optoelectronic measurement systems (Tao et al., 2012). Because no cameras are needed when working with IMUs, it is possible to perform experiments outside the laboratory, which is especially interesting for performing measurements during sports on the field (Camomilla et al., 2015). However, the three components of an IMU – accelerometer, gyroscope and magnetometer – also have disadvantages such as drifting of the gyroscope or metallic disturbances of the magnetometer. Different algorithms are used to fuse the measuring signals to keep each sensor's disadvantages as small as possible (Tao et al., 2012). The algorithms used for sensor fusion as well as the biomechanical models used for evaluating kinematic motion patterns differ between the commercial systems available on the market (Robert-Lachaine et al., 2016). Therefore, the measurement accuracy of each system needs to be evaluated and cannot be compared to another IMU-based system easily. Additionally, the performance of IMU-based systems in the main motion plane is generally superior to the performance in the other motion planes (Godwin et al., 2009) and dependent on the motion task (Robert-Lachaine et al., 2016).
Therefore, the aim of this study is to evaluate the measurement accuracy of the IMU-based measurement system MyoMotion (Noraxon U.S.A. Inc., Scottsdale, Arizona, USA) for different motion tasks compared to the gold-standard of optoelectronic measurement systems (VICON™, Oxford, UK) for lower limb joint angles in all three motion planes. So far, one validation study of this system was carried out using a previous software version with a simpler biomechanical model implemented. This study focussed on the evaluation of knee flexion angles only (Balasubramanian, 2013).

METHODS: 12 healthy subjects (7 male, 5 female, age = 26.9 ± 2.3 years, mass = 70.8 ± 13.2 kg, height = 171.9 ± 10.2 cm) participated in this study after giving written consent. Each subject performed the following tasks ten times each: level walking at a speed of 0.8 m/s, 1.1 m/s, 1.4 m/s, 1.7 m/s and 2.0 m/s ± 10 % controlled by a light barrier; sloped walking (20 % slope) at a self-selected speed, ascending and descending stairs at a self-selected speed. The motion was captured simultaneously by ten infrared cameras (VICON™, Oxford, UK, 100 Hz) with 49 reflective markers and an IMU-based system (MyoMotion, USA) consisting of 16 sensors (Figure 1), which were placed according to the manufacturer's specifications (Noraxon U.S.A. Inc., 2013).

Figure 1 Subject with attached markers and IMUs.

All trials of level walking are considered for the evaluation, and one trial per subject for sloped walking as well as ascending and descending stairs. 3D joint angles of the lower extremities were calculated by means of an anatomic-landmark-scaled Lower-Body-Model (Lund et al., 2015) from the AnyBody Modeling System (Version 6.0, AnyBody Technology, Aalborg, Denmark) for the optoelectronic data. For the IMU-based system, joint angles are calculated within the software using the sensors placed on the feet, shanks, thighs and pelvis. All further data analysis was performed using MATLAB (2016a, The MathWorks, Inc., Natick, Massachusetts, USA). Steps are defined as lasting from one touchdown to the next touchdown based on kinematic parameters (Maiwald et al., 2009) and normalised to 100 % of the gait cycle. Steps showing an offset of more than 10 % from the mean maximum knee flexion angle were excluded from the analysis. The method of Statistical Parametric Mapping (SPM) as described in Pataky et al. (2016) was used for the statistical analysis. Dependent on the data distribution, either a parametric or a non-parametric two sample t-test was performed to answer the hypothesis whether the joint angles measured by both systems differ significantly.

RESULTS: To give an example, the mean and standard deviations of all 120 steps that were performed at a walking speed of 2.0 m/s are displayed in Figure 2. As displayed in Figure 2a, the curve progression does not show a difference, but there is an RMSE between the mean values of 3.93° for the hip, 3.71° for the knee and 4.24° for the ankle. Regarding the adduction (Figure 2b), the RMSE between the mean of hip and knee is 3.81° and 3.06°, while it is 4.24° for the ankle. The RMSE between the mean rotation angles is 6.17° for the hip, 8.67° for the knee and 8.36° for the ankle. In average, the mean standard deviations are less than 5.91° for flexion, 2.49° for adduction and 4.86° for rotation for the optoelectronic system, while they are less than 6.54° for flexion, 5.86° for adduction and 11.61° for rotation for the MyoMotion system. When evaluating the mean standard deviations of single subjects, it can be seen that these differences are stronger in 6 of 12 subjects, partly up to 20° in hip and knee flexion angle.
The statistical evaluation of the 3D joint angles over all steps of all subjects per gait velocity in level walking shows a significant difference ($p > 0.05$) in the all motion planes for all joints. The ankle joint shows differences throughout the whole step, while the differences in hip and knee flexion decrease with increasing velocity.

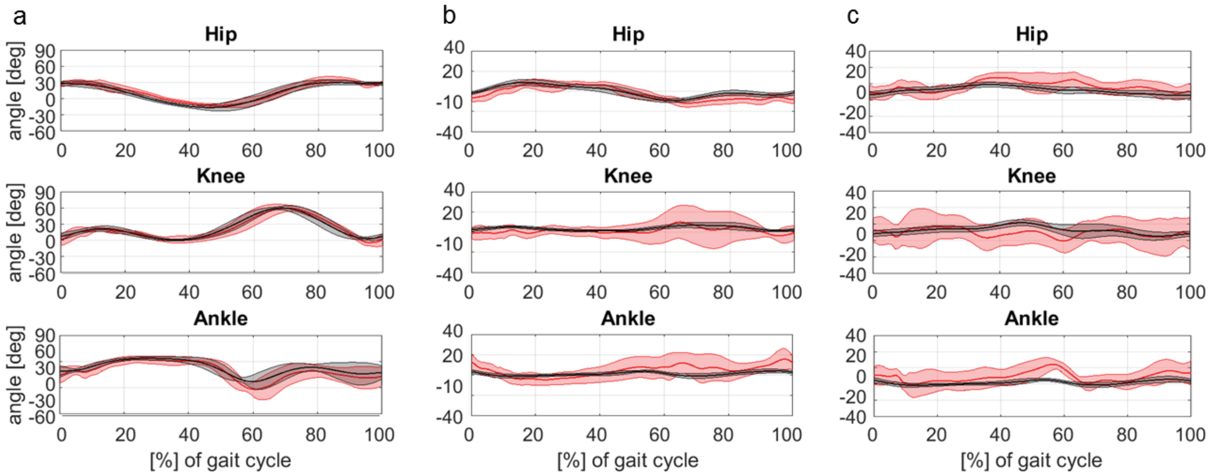

Figure 2 Mean and standard deviation of both measurement systems for (a) flexion-extension, (b) adduction-abduction, (c) internal rotation-external rotation for level walking at a speed of 2.0 m/s. The results for MyoMotion are displayed in bright red, the results for VICON in dark grey.

One trial per subject is evaluated for sloped walking. Therefore, the amount of steps is 48 for descending and 38 for ascending. Regarding this limited amount of steps, the MyoMotion system shows significant differences in the main motion plane. Ankle flexion ($p < 0.05$) differs significantly for descending. Only the results for the ankle differ significantly ($p < 0.01$) for the adduction-abduction motion. The knee rotation differs between both systems ($p < 0.01$) for both conditions, the ankle rotation differs for the ascent ($p < 0.01$).
18 and 35 steps are evaluated for ascending and descending stairs. All ankle angles differ significantly ($p < 0.05$) for both motion tasks as well as the knee rotation. Additionally, there is a difference in hip adduction ($p < 0.01$) for the descent and a difference in flexion and rotation ($p < 0.01$) for the ascent.

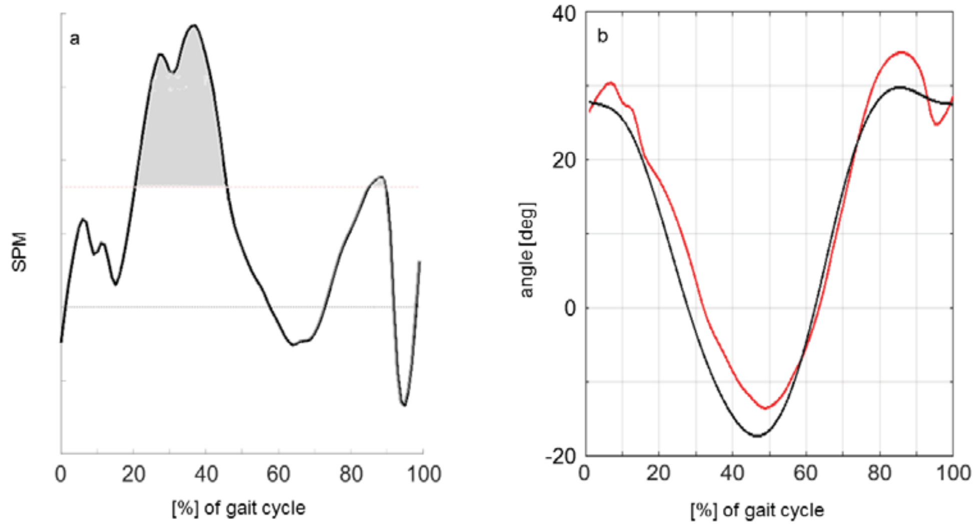

Figure 3 (a) Results of SPM, thresholded at t > 3.272. Grey areas display significant differences in mean hip flexion (p<0.05). (b) Mean values of hip flexion (VICON black, MyoMotion red) at a gait velocity of 2.0 m/s.

DISCUSSION: The results indicate that the mean joint angle progression shows a distinct offset in the main motion plane that is caused by the correction of AnyBody using its standing reference trial. After adding out the standing reference, for different subjects an offset in the main motion plane can still be determined. This offset results from the definition of the flexion

axis of the joint coordinate system in which the joint angles are defined. Due to this difference, the other two joint coordinate axes of both systems differ as well.
The high standard deviation in the MyoMotion joint angles might be explained by the repeated calibration of the IMUs. Before each trial, the subject was asked to stand in an upright calibration posture. Due to small changes in this posture, the results might differ.
In general, it can be observed that the differences in joint angles increase with increasing gait velocity. Therefore, the results regarding the descent show larger standard deviations of the MyoMotion system than those of the ascent. These differences might be explained by the higher impact in the acceleration signal during touchdown that are not filtered sufficiently. This might also explain the differences in those planes with minor motions. The evaluation of stair climbing shows differences in the hip joint angles that do not appear during the other motion tasks. Due to the higher complexity of stair climbing, this motion task might lead to difficulties in determining joint angles with the help of IMU data. To support these findings, more steps need to be evaluated.
The findings of this study support those by Godwin et al. (2009) and Robert-Lachaine et al. (2016). To distinguish between differences in angle calculation based on differences in the biomechanical model used and those based on measurement system limitations, the angles of both systems should be calculated by applying the same biomechanical model.

CONCLUSION: The results support the hypothesis that joint angles measured by an optoelectronic and an IMU-based system differ. The determination of the joint axes seems to be the main difference. Additionally, users need to consider the dependency on the gait speed velocity. While the joint angle progression in the main motion plane does not differ between both systems, the maximum joint angles differ up to 20° for the hip and 12° for the knee without the correction by the standing reference.

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