Benchmarking of a full-body inertial motion capture system for clinical gait analysis

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Abstract—In order for gait analysis to be established as part of routine clinical diagnoses, an accurate, flexible and user-friendly motion capture system is required. Commonly used optical, mechanical and acoustic systems offer acceptable accuracy and repeatability, but are often expensive and restricted to laboratory use. Inertial motion capture has seen great innovation in the last few years, but the technology is not yet considered mature enough for clinical gait analysis. In this paper we compare the kinematic reliability of inertial motion capture with optical motion capture during routine gait studies of eight able-bodied subjects. The root mean squared, RMS, and coefficient of correlation, \( R \), was used to compare data sets. Sagittal plane joint angles in the knee and hip compared very well. Corresponding transverse and frontal plane values were moderately accurate. The ankle joint angles calculated from the two systems were less accurate. This was believed to be due to the use of different rotation axis orientations used for calculation of angular rotations.

I. INTRODUCTION

In the last decade several papers have surfaced on the use of inertial motion capture (IMC) for ambulatory motion analysis [1]. This is largely due to recent advances in micro-electromechanical system (MEMS) technologies allowing for the development of small and inexpensive sensors such as accelerometers, gyroscopes and magnetometers which form the key components of inertial motion sensor units [2,3]. Measurements from these sensors are combined in sensor-fusion and anti-magnetic-disturbance algorithms to produce accurate and virtually drift free motion tracking [1]. Simon [2] recently assessed the benefits and limitations of the application of gait analysis to clinical problems. He found that gait analysis has proven very useful in evaluating the extent of neuromuscular-skeletal conditions and in determining the appropriate intervention or surgical treatment required. He found, however, that several factors limit the attractiveness of gait analysis in patient care. Factors include expensive laboratory personnel and maintenance costs as well as low patient throughput rates (long test setup times and long waiting periods for test reports). Doubts about variability, inaccuracy and lack of reproducibility caused by technical factors, test subject factors and clinical interpretation factors also play a role [2].

With evident cost and flexibility advantages over alternative, laboratory-based optical, acoustic and mechanical systems, it would seem that IMC offers solutions to several of these hampering factors. We therefore stand to ask why IMC is still largely underutilized for clinical gait analysis. A possible answer lies in the maturity of the technology, and therefore the lack of confidence in its measuring accuracy. This study aims to address this hindrance by investigating the validity and repeatability of results obtained from a commercially available, full-body IMC system, by comparing them to results obtained from a proven and commonly used optical motion capture (OMC) system. Various other studies have been conducted to compare these technologies in specific scenarios [4]-[8], but the authors have found no evidence of such a comparison done for routine gait analysis. This paper shows the results obtained from conducting gait studies of eight able-bodied subjects using both methods simultaneously. Research is continuing to compare within-day and between-day repeatability.

II. DATA ACQUISITION

Full three-dimensional (3D) kinematic ambulatory gait analysis was conducted for eight able-bodied male subjects using simultaneous IMC and OMC. Subjects were chosen on the conditions that they had no history of any musculoskeletal impediment or any injury which might have affected gait measurements.

A. Motion capture

IMC was conducted using the full-body motion capture suit (MOVEN inertial motion capture system, Xsens technologies B.V., Enschede, the Netherlands). The MOVEN measures three-dimensional (six-degree-of-freedom) position and orientation of body segments in a global coordinate system by using 16 inertial sensor units hidden in a full-body Lycra suit. Each sensor unit contains 3D gyroscopes, 3D accelerometers, 3D magnetometers and a temperature compensation sensor. Gyroscopes measure angular velocity which is integrated over time to find segment orientation (relative to an initial orientation). Accelerometers measure linear acceleration which, after removing the gravity component, is twice integrated to find segment position (relative to an initial position). These parameters, along with magnetometer and temperature information, are fed into a Kalman filter which continuously updates to correct drift and joint position uncertainties. Sensor modules are positioned on the head, shoulders, pelvis, upper and lower arm, upper and lower leg, hands and feet (Fig.1.) [1].
Sensor data is sent from two wireless master modules on the lower back to wireless receivers plugged into a laptop computer. Data capture and visualization was done using Moven Studio V2.0 (Xsens technologies B.V., Enschede, the Netherlands). Inertial data was recorded at 100 Hz. To analyze the validity of kinematic gait parameters measured by the IMC system, a reference system was necessary. An eight camera OMC system (VICON, Oxford Metrics Ltd.) was used as the reference standard for this study, as it is currently the most popular and accepted system for clinical gait analysis. 35 reflective markers were stuck on the suit according to the “Gollum” marker model. This includes markers on the bilateral forehead, bilateral mastoid processes, single Cervical spine process 7, single thoracic spine process 10, single sternal notch, single xiphisternum, bilateral acromioclavicular joints, bilateral scapula angles, bilateral 6th costochondral joints, bilateral mid-lateral humerus, bilateral humerus epicondyle, bilateral, styloid process, bilateral ulna process, bilateral third dorsal metacarpal, bilateral anterior superior iliac spines sacrum, bilateral lateral mid thigh, bilateral femoral lateral condyle, bilateral lateral mid tibia, bilateral lateral malleoli, bilateral calcaneenii, and the bilateral first metatarsophalangeal joint. VICON data was collected (at 250 Hz) and processed using BodyBuilder software (VICON, Oxford Metrics Ltd.).

B. Procedure
Subjects were asked to walk at five different speeds (slow walk, normal walk, fast walk, jog, run and sprint) along a 7 m walkway. Three trials were collected per subject for each of the five speeds. Static calibration (first for the VICON and then for the MOVEN) was done shortly before the start of each trial to minimize the effects of inertial sensor drift.

III. DATA ANALYSIS
Moven Studio collects and stores data in a proprietary file format (*.mvn). This data was then converted (using Moven Studio) to a .mvnx file format which contains three-dimensional position values (Cartesian coordinates in a global coordinate system) and three-dimensional orientation values in the form of unit quaternions [1] at each time frame for each of the 23 body segments created in the Moven Studio body model. Lower body kinematic joint angles were found by calculating the joint orientation of the distal segment with respect to the proximal segment (Fig.2). This is achieved through a quaternion multiplication of the complex conjugate of the proximal segment quaternion $GBq_U$ and the distal segment quaternion $GBq_L$ [1]:

$$Bq_{UL} = GBq_U \otimes GBq_L$$

Angles were then converted to Euler angles [9]:

$$\begin{bmatrix}
\phi \\
\theta \\
\psi
\end{bmatrix} = \begin{bmatrix}
\text{atan2} (2q_2q_3 + 2q_0q_1, q_3^2 - q_2^2 - q_1^2 + q_0^2) \\
- \text{asin} (2q_2q_3 - 2q_0q_2) \\
\text{atan2} (2q_1q_2 + 2q_0q_3, q_1^2 + q_0^2 - q_2^2 - q_3^2)
\end{bmatrix}$$

where $\text{atan2}$ and $\text{asin}$ are MATLAB (Mathworks, Natick, MA) commands for the four quadrant inverse tangent and the inverse sine respectively.
To compare the data collected from inertial sensors with that of the optical system, the corresponding data curves (of a single gait cycle) were fitted on top of each other and the coefficient of correlation, \( R \) as given by [11] was calculated:

\[
R = \frac{n \sum xy - (\sum x)(\sum y)}{\sqrt{n(\sum x^2) - (\sum x)^2} \sqrt{n(\sum y^2) - (\sum y)^2}}
\]

where \( x \) and \( y \) are data points at \( n \) time intervals of the data sets to be compared.

A constant bias error was found when fitting the curves on each other. This could have been because the calibration processes used by the IMC and OMC systems differ and were not conducted simultaneously. An optimization algorithm was used in which the OMC data was shifted vertically over the IMC data (by one degree per iteration) until the sum of the squared errors at each time step was at its minimum. The effect of this bias error was evaluated by calculating mean and STD values for the RMS in degrees, with and without bias removal.

IV. RESULTS

Fig. 3a-c show the joint angle comparison of one subject for a normal walking speed (0.5-3 m/s). Mean RMS and \( R \) values and the related STD values are given in Table I and Table II respectively. Table II also shows the \( R \) values which correspond to those represented in Fig. 3. For all eight subjects, the sagittal plane measurements of the hip and knee measured by IMC closely resembled those of OMC with an average \( R \) value of 0.94 and 0.89 respectively. The corresponding values for dorsiflexion/plantar flexion of the ankle does not compare well, with \( R \) values averaging 0.08.

V. DISCUSSION

Besides skin artifacts, common in motion capture [12], the Lycra suit caused additional anatomical model landmark calculation errors because the markers were adhered to the suit. In a study by Dejnabadi et. al. [5], a more controlled comparative test was conducted where inertial markers were fastened securely with straps to the thigh and shin and which only looked at the knee flexion angles. They found \( R \) values in excess of 0.99. This leads to the conclusion that errors due to the Lycra suit were a significant contributor to errors in our study.

The discrepancy in ankle angles is believed to be caused by the difference in calculation of the rotation axes. The shape of ankle flexion curves calculated from IMC data resembles those found in literature [13,14]. Better securing of the IMC foot sensor and the use of a different OMC marker set may resolve this discrepancy. The OMC gait laboratory used was limited to a short capture distance. This meant that at higher speeds (>3 m/s) a full stride could not be captured for both legs. Some data loss caused by shadowing was also experienced. However these errors were considered negligible.

Fig. 3. Left side joint angles in the sagital, frontal and transverse planes are given for the a) hip, b) knee and c) ankle for a normal walk speed for one of the tested subjects.
VI. CONCLUSION

At lower walking speeds the reliability of data measured using the IMC system is comparable with that of the OMC system. Sagittal plane angles of the hip and knee joints are reasonably accurate while angles in the transverse and sagittal planes are marginally comparable. More tests are required to determine the shortcomings in calculated ankle angles. As is, the IMC allows quick and easy visualization of gait measurements in any environment, but had limited accuracy in some measurements due to the relative motion between the skin and the Lycra suit. A major advantage of the IMC is its wireless capabilities, its ability to record continuous data for numerous gait cycles, its quick set-up time (significantly less than the OMC), and comparatively lower cost.

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VII. FUTURE WORK

Further research is currently underway to investigate the repeatability of the IMC system for within and between day gait trials. Suit artifacts will be explored by attaching the inertial sensors directly onto the skin of the test subjects and comparing the calculated values of CMC to those computed using the Lycra suit. Preliminary results show that within-day repeatability is enhanced by strapping sensors directly to the skin.

The IMC system is only capable of measuring kinematic gait data. An interesting study may be to incorporate pressure sensors into the feet of such a suit to calculate kinetic parameters and act as a replacement to the lab-based force plates. With the addition of EMG electrodes inside the suit, a fully functional, fully mobile gait measuring suit could help gait analysis in becoming a part of routing medical check-ups. Future research may lead to the development of a single suit capable of simultaneously measuring a vast number of physiological parameters including: ECG, EEG, respiratory sounds, oxygen saturation, body temperature, kinematic and kinetic gait patterns, EMG and many more.

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REFERENCES