A Study on Estimation of Planar Gait Kinematics using Minimal Inertial Measurement Units And Inverse Kinematics

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Abstract— **This paper describes a preliminary study of using four inertial measurement units (IMUs) attached to the heel and pelvis to estimate the joint angles of normal subjects during walking. The IMU, consisting of a 3-D accelerometer and gyroscope, is used to estimate the planar displacement of the heel and pelvis and the angular change of heel in one gait cycle. We then model the gait as a planar 3R serial chain and solve its inverse kinematics by using such information. The results are validated by comparing the estimated joint angles of lower limbs (i.e. hip, knee and ankle angles) with an optical motion capture system. This study can benefit the future research on conducting complete lower limbs kinematics analysis with minimal and unobtrusive wearable sensors.**

I. INTRODUCTION

In recent years, wearable sensors such as inertial measurement unit (IMU) have been widely used in gait analysis. Due to the advancements in microelectronics technologies, IMUs are small, lightweight and low-cost. They are capable of capturing body movement unobtrusively and allow kinematic measurements to be monitored over extended space and time period. Compared to the traditional optical motion capture system, wearable sensors facilitate real-time outdoor data collection to monitor human gait in a more natural way.

IMUs have been utilized for capturing many gait related measurements, such as timing of gait events (e.g. heel contact and toe off), stride length, walking velocity, body segment orientation and position [1], [2], [3]. However, only a few studies used IMUs for obtaining the lower limb kinematics. The difficulties are probably due to the relative large numbers of IMUs required for a complete kinematic analysis of the lower limb. Usually, it requires at least two IMUs for calculating the relative motion between two consecutive lower limb segments. For instance, Dejnabadi et al. [4] used two IMU sensors, each at the shank and thigh, for obtaining accurate knee angle measurements. Mayagoitia et al. [5] used four uniaxial accelerometers and one gyroscope per body segment to obtain the kinematics of lower extremity. Watanabe et al. [6] and Meng et al. [7] used seven sensors bilaterally for obtaining the hip, knee and ankle angle measurements.

The number of IMUs required increases the complexity and cost of the data collecting system. It may be obtrusive and interfere with the human movement. It may also reduce the level of user compliance [8]. A way to solve this problem is to collect data from a smaller number of sensors and estimate these kinematic measurements rather than measuring them directly.

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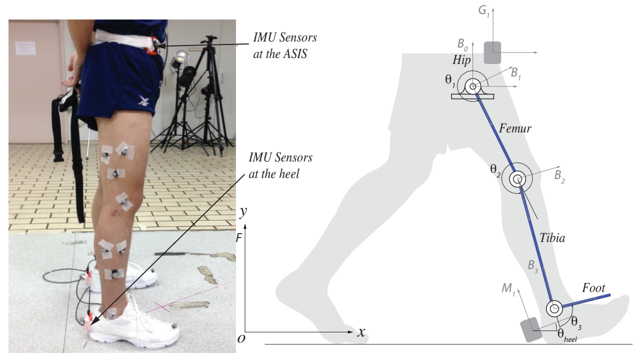

Fig. 1. IMUs and reflective markers placement and the schematic drawing of our gait model. The joint angles θ_1, θ_2 and θ_3 denotes the lower limb parameters during walking and are used in our inverse kinematics analysis.

This paper describes a novel approach for calculating hip, knee and ankle angles during normal walking with a reduced number of IMUs. Instead of measuring the joint angles directly, we use the IMUs to obtain critical relative limb motions and estimate the joint angles in an indirect manner. We model the gait as a planar 3R serial chain and estimate the displacement and orientation of foot and pelvis by the IMUs attached at the heel and Anterior Superior Iliac Spine (ASIS). These positions are used in conjunction with geometric constraints to solve the inverse kinematics of a planar 3R serial chain to obtain the hip, knee and ankle joint angles on sagittal plane. This helps us to reduce the required number of sensors to only four IMUs.

II. METHOD

A. Joint Angle Estimation System

The measurement system consists of four IMUs (Xsens MTx). Two IMUs were strapped to the heel of a subject's shoes and the another two were attached to the subject's medial and lateral ASIS, see Fig.1. To estimate the joint angles, we use a kinematic gait model as shown in Fig. 1. This model consists of 4 rigid links, which correspond to the Hip, Femur, Tibia and Foot, and 3 revolute joints that connect between these links. The general methodology was to estimate the joint parameters of this serial chain by inverse kinematics, based on the positions estimated by the IMUs located at the heel (M_i) and ASIS (G_i). These positions were determined from the angular velocity and acceleration measured in the corresponding IMU.

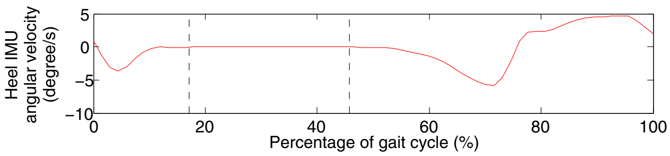

Fig. 2. Foot angular velocity measured by the heel IMU in one gait cycle. The dash lines indicate when the magnitude of foot angular velocity is approximately zero (<0.05 degree/s) and foot comes to a stop.

There are three major challenges in using IMU for determining these positions. Firstly, there will be an integration drift due to data noise. Secondly, zero offset will exist and needs to be corrected as we integrate the measured IMU signals. Thirdly, IMU only measures local displacements and a global relationship between the ASIS and heel position needs to be established. To tackle these obstacles, some assumptions have been made based on the geometrical constraints during walking. They are described in the next few sections.

B. Estimation of the Heel IMU Displacement

We segmented the walking motion into a series of gait cycles before computing its displacements. The timing of each gait cycle was determined by the minimal vertical displacement of the heel IMU. To reduce integration drift, we assumed that the value of $a_{heel,x}(t)$ and $a_{heel,y}(t)$ during foot flat were zero. This phase occurred when the slight sliding motion of the foot came to a stop, right after the initial heel contact [9]. The timing of this phase was determined from the foot angular velocity when it was approximately zero (<0.05 degree/s). As shown in Fig. 2, this corresponded to the region between the two vertical dotted lines. The associated velocities $v_{heel,x}(t)$ and $v_{heel,y}(t)$ within each gait cycle were calculated using,

$$\begin{aligned} v_{heel,x}(t) &= \int_0^t a_{heel,x}(\tau)d\tau + v_{heel,x}(0), \\ v_{heel,y}(t) &= \int_0^t a_{heel,y}(\tau)d\tau + v_{heel,y}(0), \end{aligned} \tag{1}$$

where the $v_{heel,x}(0)$ and $v_{heel,y}(0)$ were determined as the magnitude of local minima of $\int_0^T a_{heel,x}(\tau)d\tau$ and $\int_0^T a_{heel,y}(\tau)d\tau$ respectively over a gait cycle of period T. Again, the values of $v_{heel,x}(t)$ and $v_{heel,y}(t)$ during foot flat were set to zero to further reduce the drifting effect.

Now, the associated heel displacements $d_{heel,x}(t)$ and $d_{heel,y}(t)$ can be computed using,

$$\begin{aligned} d_{heel,x}(t) &= \int_0^t v_{heel,x}(\tau)d\tau + d_{heel,x}(0), \\ d_{heel,y}(t) &= \int_0^t v_{heel,y}(\tau)d\tau + d_{heel,y}(0), \end{aligned} \tag{2}$$

where $d_{heel,x}(0)$ and $d_{heel,y}(0)$ were set to zero since we decided to set this particular position as the global measurement frame F.

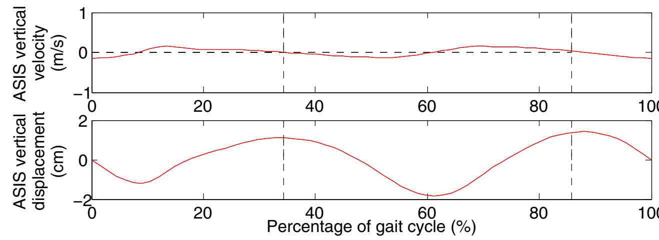

Fig. 3. The ASIS vertical velocity and displacement within one gait cycle. The ASIS vertical velocity equals to zero at the time of local maxima of vertical displacement.

C. Estimation of the ASIS IMU Displacement

At the ASIS, the magnitude of its acceleration is smaller than that of heel during walking. Therefore the integration is more prone to integration drift. To correct for this, we implemented the iterative integration method proposed by Bamberg [1] to determine both its velocity and displacement.

For each Bamberg iteration, the velocity of the ASIS IMU was integrated similar to Eq. (1). We assumed that the ASIS initial horizontal velocity $v_{ASIS,x}(0)$ was equal to the mean walking velocity of the gait cycle. This was determined using the heel horizontal displacement $d_{heel,x}(T)$ and the time T took,

$$v_{ASIS,x}(0) = \frac{d_{heel,x}(T)}{T}. \tag{3}$$

While for the ASIS initial vertical velocity $v_{ASIS,y}(0)$, we computed it using a two step procedure. We first integrated $v_{ASIS,y}(t)$ disregarding its initial vertical velocity to locate the time instance t_{max} when the local maxima of its vertical displacement occurred. Then, we determined its initial velocity $v_{ASIS,y}(0)$ using,

$$v_{ASIS,y}(0) = -v_{ASIS,y}(t_{max}). \tag{4}$$

We then computed the associated ASIS displacements $d_{ASIS,x}(t)$ and $d_{ASIS,y}(t)$ similar to Eq. (2) using Bamberg iteration. The final result is as shown in Fig. 3.

A crucial part of our measurement system was to establish the initial relationship between the ASIS and heel IMU positions for our inverse kinematic analysis. We made use of the geometrical property during contralateral heel contact to relate this. Consider a biomechanical model as shown in Fig. 4 with heel contact at a particular time t_0 and its following contralateral heel contact at time t_1. From the attached IMUs at the heel and ASIS, we can obtain the horizontal distance travelled by the hip d_1 and ankle d_2 during this particular time interval. This allowed us to express the initial ASIS horizontal displacement, $d_{ASIS,x}(0)$ in terms of the following variable x_1 as follows:

$$d_{ASIS,x}(0) = d_2 - d_1 - x_1. \tag{5}$$

We assumed the mean ASIS vertical displacement in one gait cycle equals to the ASIS height measured at the upright standing (H) since the ASIS vertical displacement can be approximated as a sinusoidal oscillation [10]. Therefore the

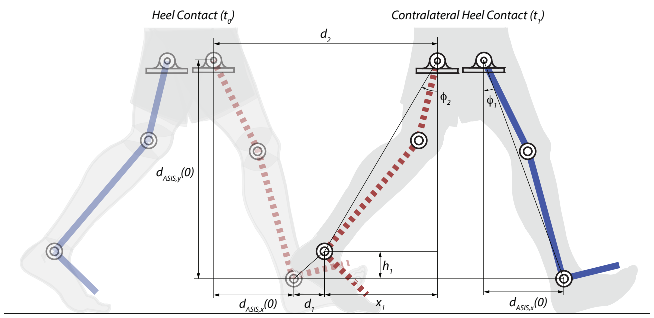

Fig. 4. A biomechanical model indicating relative displacement of hip and ankle at the time of heel contact of the stance leg (red dotted) and the following heel contact of contralateral leg (blue).

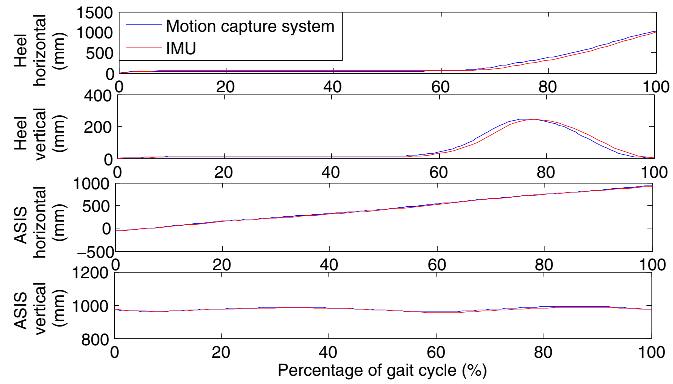

Fig. 5. The ASIS and heel displacement on the sagittal plane as estimated by the IMU, compared with the corresponding marker displacement from a motion capture system.

initial ASIS vertical displacement, $d_{ASIS,y}(0)$ was calculated as follows:

$$d_{ASIS,y}(0) = H - \frac{\int_0^T v_{ASIS,y}(\tau)d\tau}{T} \tag{6}$$

The angles of stance and swing leg relative to the vertical axis (Φ_1 and Φ_2) are equal, according to [11]. Therefore, we can relate $d_{ASIS,y}(0)$ to the vertical distance travelled by the ankle h_1 and variable x_1 using similar triangles as follows:

$$\frac{d_{ASIS,x}(0)}{x_1} = \frac{d_{ASIS,y}(0)}{d_{ASIS,y}(0) - h_1}. \tag{7}$$

Solving these equations yielded the ASIS initial horizontal displacement.

D. Estimation of Gait Parameters

Once the displacements of the IMUs are established, we can compute the joint parameters of the gait using inverse kinematics. We determined the knee joint angle for a given position of the foot using,

$$\theta_2(t) = \arccos \frac{d_x(t)^2 + d_y(t)^2 - a_{12}^2 - a_{23}^2}{2a_{12}a_{23}}, \tag{8}$$

where $d_x(t) = d_{heel,x}(t) - d_{ASIS,x}(t)$ and $d_y(t) = d_{heel,y}(t) - d_{ASIS,y}(t)$. a_{12} and a_{23} corresponded to the anthropometric length of the Femur and Tibia respectively.

We computed the hip angle using,

$$\theta_1(t) = \arctan \frac{d_y(t)}{d_x(t)} - \arctan \frac{a_{23} \sin \theta_2(t)}{a_{12} + a_{23} \cos \theta_2(t)}. \tag{9}$$

In order to achieve smooth results, a cubic smoothing spline curve fitting was used to generate the angular changing profile for these two joint parameters during a gait cycle.

Finally, the foot angle $\theta_3(t)$ was obtained from the fact that the rotation of the end-link is $\theta_{Heel} = \theta_1 + \theta_2 + \theta_3$, which yielded,

$$\theta_3(t) = \theta_{heel}(t) - \theta_1(t) - \theta_2(t). \tag{10}$$

III. RESULTS

An eight-camera motion capture system (Motion Analysis Eagle System) was used simultaneously for experimental verification. Twenty-two reflective markers were placed bilaterally at ASIS, thigh segment, knee joint, shank segment, ankle, heel and toe (see Fig.1).

Data were collected from one subject for this preliminary study. Before walking, the subject was asked to stand still in the upright posture, facing the walking direction. And the orientations of the IMUs were initialised to zero at this posture. The subject was instructed to walk at self-selected speed on a linear walkway (12m $\times$ 1.5m) covered by vinyl tiles with the IMUs and the motion capture system collecting data simultaneously. The collected data were filtered using a second order, zero-phase-lag, low-pass Butterworth filter with the cut-off frequency set at 10Hz.

The anthropometric data of the subject were estimated by measuring from the reflective markers during the quiet stand (i.e. stand still in the upright posture). The hip joint center was estimated to be at 24% of pelvic width posteriorly, and 30% of pelvic width inferiorly relative to the ASIS [12]. The ASIS height (H) was the vertical reading of ASIS markers measured from the motion capture system. It was estimated to have a value of $H = 1025mm$. The Femur length was estimated as the distance between hip joint center and knee marker. The Tibia length was estimated as the distance between knee marker and ankle marker. These yielded the mean value of $a_{12} = 430mm$ and $a_{23} = 420mm$.

Fig. 5 shows the results of the displacement estimated by the IMUs on the sagittal plane during a gait cycle. They were compared with the corresponding kinematic data calculated based on the motion capture system. Fig. 6 shows the lower limb joint angles on sagittal plane as estimated by our model in comparison with the data obtained by the motion capture system.

IV. DISCUSSION

The results show that the lower limb joint angles can be estimated accurately by our model. As shown in the joint angle profiles (Fig. 6), the differences between the estimated

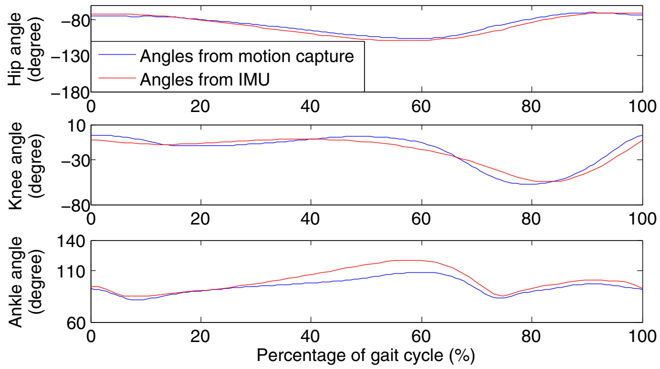

Fig. 6. The lower limb joint angles on the sagittal plane as estimated by the IMU, compared with the angles obtained from a motion capture system

angles and the angles based on the motion capture system are in general less than 10 degrees. This error is within an acceptable range compared to other researchers. For example, the errors of estimated angles reported in [6] were within the similar range. Bakhshi et al. [13] also reported an average error of knee angle estimation at approximately 2.4 degree (standard deviation=13.30) during a combined movement of gait and squatting.

We found that the accuracy of our measurement system relies much on the precision of the displacement estimated by the IMU sensors at ASIS and foot. However, there is always an integration drift due to data noise affecting performance. We have addressed this challenge by implementing biomechanical constraint at the heel during walking. For each gait cycle, there is a phase when the heel remains flat to the ground and we assume both its acceleration and velocity are zero during this period. This allows us to reduce drifting due to noise.

However, for the ASIS acceleration, it is difficult to find such biomechanical constraint. To our knowledge, no study has been provided in this regard. Besides, the magnitude of its acceleration is approximately 3 times smaller than that of heel during walking. Therefore the integration is more prone to integration drift. We implement an iterative integration method proposed by previous study to minimize the drift [1]. The comparison between the estimated displacement and that obtained from the corresponding reflective markers (Fig. 5) shows this approach can reduce the drift effect adequately.

The sources of errors in our angle joints estimation are twofold. First, despite the effort we made, there are still some errors for the heel and ASIS position estimation. The magnitudes of the errors are within 10 mm range and are smaller than those in previous research, such as [1] and [2]. However, this error might lead to an inaccurate estimation of the angles. Another source of error is likely due to the simplified planar model. The scale of joint angles change of lower limbs in the sagittal plane is much larger than that in other body plane. Therefore to simplify the application, a 2D model is implemented. However, there is a slight displacement of the lower limbs in the mediolateral direction due to the pelvis rotation. Therefore, in reality, the link lengths between the joints are changing slightly due to this mechanism. The 2D model didn't register this change.

V. CONCLUSION AND FUTURE WORK

This study demonstrates that the lower limbs joints during gait can be adequately estimated by only four IMU sensors. The joint angles estimated are accurate with only small errors within acceptable range. The errors due to the inaccurate heel and ASIS position estimation can be minimized by further improvement on the integration method. This study has revealed that the integration error can be reduced by implementing geometric biomechanical constraint. For future work, more experiments will be carried out with more subjects and gait trials to statistically analyse and verify our model performance.

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