An Inertial Sensor Based System for Real-Time Biomechanical Analysis during Running

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***Abstract*—Running is a popular form of physical exercise. However, runners often experience a high number of running-related injuries. This is partially due to the fact that runners are lack of knowledge about their running postures during running. This paper describes a preliminary study of using an Inertial Measurement Unit (IMU) based wearable system for real-time biomechanical analysis during Running. This system consisted four IMU sensors. Two were strapped to the toes of a subject's shoes and another two were attached to the subject's medial and lateral ASIS. The foot and pelvis positions were estimated by the IMU acceleration and a linear regression model. The lower limb joint angles were estimated by combining these position information with a planar 3R serial chain and solving its inverse kinematics. The results show that joint angles of lower limbs (i.e. hip, knee and ankle angles) can be accurately estimated by this wearable system. This study can benefit the future research on conducting complete lower limbs kinematics analysis with minimal and unobtrusive wearable sensors and provide real-time feed back during running exercise.**

***Index Terms*—biomechanics, running, wearable system, joint angles**

I. Introduction

Running is a popular form of physical exercise. There are large running polulation around the world. For instance, there are more than 40 million runners in the US [1]. However, runners often experience a high number of running-related injuries. The yearly incidence rate for running injuries were reportedly between 37% and 56% [2]. Each year, approximately 25%-50% runners sustain an injury that is severe enough to cause a change in practice or performance [3]. The running injury rate is especially high for novice runners [4], [5]. Such increased incidence of running-induced injury is partially due to that novice runners are lack of knowledge about the pathophysiology and biomechanics of chronic running injuries [6].

Although tremendous amount of scientific and medical research has been carried out to study and prevent running injuries, and many runners have gradually taken precautions about running injuries, there are still huge gap in terms of allowing the results of scientific research to be beneficial towards common runners and prevent them from serious injuries. This gap is partially due to the fact that quantitative biomechanical analysis was often restricted to laboratory environment. Traditional biomechanical analysis facilities such as the optical motion capture system and force plate are limited in capture volume [7].

The recent advancements in microelectronics technologies have improved the wearable sensors to be used for biomechanical analysis. Wearable sensors have inherent advantages over the conventional laboratory-based facilities. They are small, lightweight, and capable of monitoring human movement over extended space and time, thus allow the biomechanical analysis to be conducted unobtrusively and in outdoor settings.

One of the most commonly used wearable sensors for biomechanical analysis is the Inertial Measurement Units (IMUs) [8]. IMUs often combine accelerometers and gyroscopes. Some might also include magnetometers. These sensors are often integrated into a compact structure that can be easily attached to the human body. IMUs can provide accurate acceleration and orientation data, which makes them very suitable for human motion tracking and analysis. In fact, IMUs have been widely used for gait-related research and applications. For instance, IMUs have been used to detect various gait events such as the heel contact and toe-off under both normal walking and pathological walking [9]-[11]. IMUs have also been used to estimate spatiotemporal parameters such as the stride length, cadence, and walking velocity [12]-[15]. But more importantly, IMUs can be used to estimate body kinematics. For example, Shoe-attached IMUs have been used to estimate the heel displacement [16]. Similarly, Mariani *et al.* [17] used a foot-worn IMU system to estimate the heel and toe trajectory. Sabatini *et al.* [18] implemented IMUs to calculte the foot velocity and orientation. Liu et al. placed three IMUs at the lower extemities and estimated the body segment orientations.

IMUs have also be used to etimate lower limb joint angles during locomotion. For instance Findlow *et al.* estimated the hip, knee and ankle joints bilaterally with only four IMUs attached to the shank and foot. Pietro *et al.* used four IMUs, attached at the sacrum, thigh, tibia and tarsal bones unilaterally, to estimate the three lower limb joint angles of one leg.

In an early study [19], we proposed a novel approach that we used IMUs attached at the Anterior Superior Iliac

Manuscript received October 26, 2016; revised April 26, 2017.

doi: 10.18178/jomb.6.1.1-5

Spine (ASIS) and shoes to estimate the joint angles of hip, knee and ankle during walking. We modeled the lower legs as a planar 3R (R denotes revolute) model and estimate the joint parameters using inverse kinematics. This approach showed promising results for joint angle estimation during normal walking. In the presenting study, we aimed to implement this model for estimating the joint angles of lower limbs during running. Based on that, we proposed to a low cost IMU sensor based wearble system that can estimate the lower limb joint angle during running.

II. METHOD

A. IMU Based Wearable System

This IMU sensor based system consisted four IMU sensors. Two IMUs were strapped to the toe of a subject's shoes and another two IMUs were attached to the subject's medial and lateral ASI, as schematically shown in Fig. 1. Each IMU sensor was customized and packaged in a 3D printed box with the size of 45mm (H) x 25mm (W) x 13mm (Thickness). This compact design, as shown in Fig. 2, allows the IMUs be easily attached to the human body.

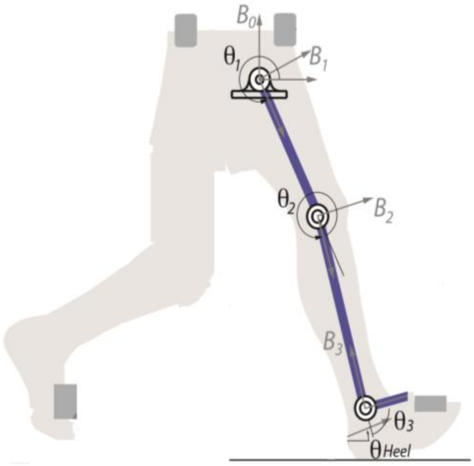

Figure 1. Low limb kinematic model (running) and sensor attachment

Each IMU sensor composed a MPU9250 composite chip (InvenSense Inc., CA, US). It integrated a 3D accelerometer, a 3D gyroscope and a 3D magnetometer. It has an on-board Digital Motion Processor (DMP) and a built-in sensor fusion algorithm which can generate accurate sensor orientation (in local frame) and gravity-free-free-acceleration of the sensor. Besides, a micro-computer PIC (nRF51822, Nordic, Sweden) was used to record and sample the IMU data and transmit the data to a PC-end receiver via Bluetooth. The MPU9250 and micro-computer PIC were soldered together (Fig. 2). The sampling rate was set to 50 Hz. The biomechanical analysis was done by a customized LABVIEW (version 8.2.0, National Instrument, US) script.

B. Joint Angle Estimation

A kinematic gait model similar to [19] was developed to estimate the lower limb joint angles. As shown in Fig. 1, this model consists 3 revolute joints (corresponding to hip, knee and ankle joints) that connect 4 body segments (pelvis, femur, Tibia and foot). The lengths of the body segments were measured in advance, so that the 2D joint angles of this serial chain can be estimated by knowing the positions of each end of this chain (i.e. the pelvis and foot position).

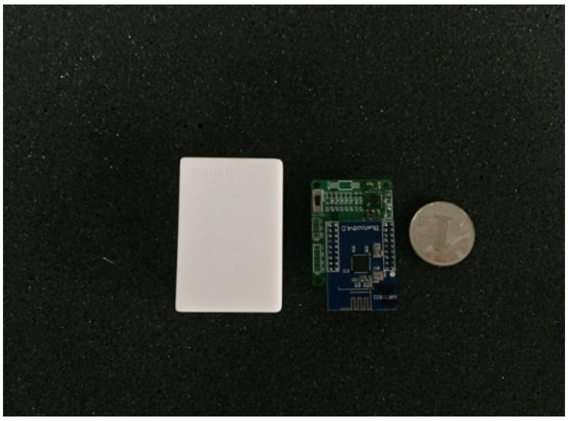

Figure 2. The customized IMU sensor

In our previous study [19], the positions of the foot and pelvis during walking were estimated by integrating the acceleration data obtained from the IMUs attached to the corresponding body segment respectively. To address the integration drift due to data noise and zero offset, we made some assumptions based on the geometrical constraints during walking [19]. However, such assumption may no longer be valid during running. Therefore, in the presenting study, the positions of the foot and pelvis were calculated by a linear regression model (LRM).

To construct the LRM, one volunteer was invited to conduct a running trial. The volunteer was asked to wear our IMU sensor based system. Meanwhile, an eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used to capture his kinematic data simultaneously. VICON's lower body Plug-in-Gait marker set was used (Fig. 3) to capture the kinematic data. VICON Nexus software was used for marker labeling, modeling and analysis with a frequency of 50 Hz. The output from the motion capture system included all the marker trajectories and the lower limb joint angles. These data were considered as the ground truth for the LRM model construction and validation.

Before the data collection, the volunteer was asked to stand still in the upright posture, facing the walking direction. And the orientations of the IMUs were initialized to zero at this posture. The anthropometric data of the volunteer, such as the leg lengths, were estimated by measuring from the reflective markers when he was standing 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 LASIS (left-hand side) or RASIS (right-hand side) [20]. The ASIS height was mean of the vertical reading of LASIS and RASIS markers. It was estimated to have a value of 1020 mm. The Femur length was estimated as the distance between hip joint center and knee marker. The mean value was 448 mm. The Tibia length was estimated as the distance between knee marker and ankle marker. The mean value was 442 mm.

At the beginning of data collection, the volunteer was asked to jump for 3 times. The jumping was used later for synchronizing the VICON data and IMU data. After that, the volunteer was instructed to run at self-selected speed on a treadmill for 3 minutes. IMU system and the motion capture system were 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.

Running was segmented into a series of running cycles. It starts from the foot making contact with the ground and ends at the same foot making another contact. This was automatically detected by obtaining the local maximum of the sum of the square of 3D acceleration obtained from the shoe-attached IMU. Within each running cycle, a LRM was used to weight the IMU acceleration to match the foot and pelvis positon captured by the motion capture system, as:

$$b_1 a_i + b_2 a_{i+1} + b_3 a_{i+2} + b_4 t_i^2 + b_5 t_{i+1}^2 + b_6 t_{i+2}^2 + b_7 = D \quad (1)$$

where b_{1-6} were the model coefficients, and b_7 was the shifting constant. a_i was the acceleration time series, t_i was the corresponding time. D c the pelvis and foot position captured by the motion capture system. A total number of 20 running cycles were used to estimate the model parameters.

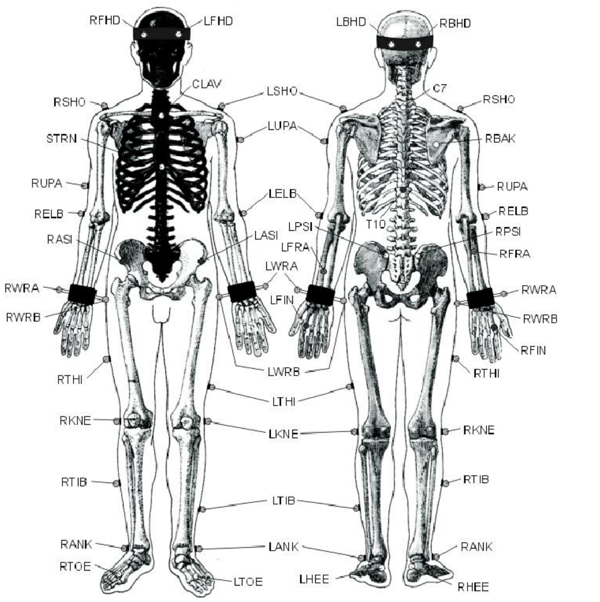

Figure 3. Marker placement (VICON Plug-in-Gait Model).

Once the positions of foot and pelvis were established, the lower limb joint angles can be determined as [21]:

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

$$\theta_1(t) = \arctan \frac{d_y(t)}{d_x(t)} - \arctan \frac{a_{23}\sin\theta_2(t)}{a_{12} + a_{23}\cos\theta_2(t)} \quad (3)$$

$$\theta_3(t) = \theta_{heel}(t) - \theta_1(t) - \theta_2(t) \quad (4)$$

where $d_x(t)$ and $d_y(t)$ corresponded to the distance between the foot segment and pelvis segment, which were calculated by their positions. a_{12} and a_{23} corresponded to the anthropometric length of the Femur and Tibia respectively. Theta 1-3 correspond to hip, knee and ankle joint angles respectively. In order to achieve smooth results, a cubic smoothing spline curve fitting was used to generate the angular changing profile for all the joint angles.

III. Results

A. IMU Sensor Output

IMU sensor output in a time window of 5 seconds are shown in Fig. 4. Both the acceleration and orientation data are in cyclic form. Peak acceleration can be clearly observed in foot IMU acceleration. Such peaks were used to segment the running motion into a series running cycles. Fig. 5 shows the typical IMU output within one running cycle.

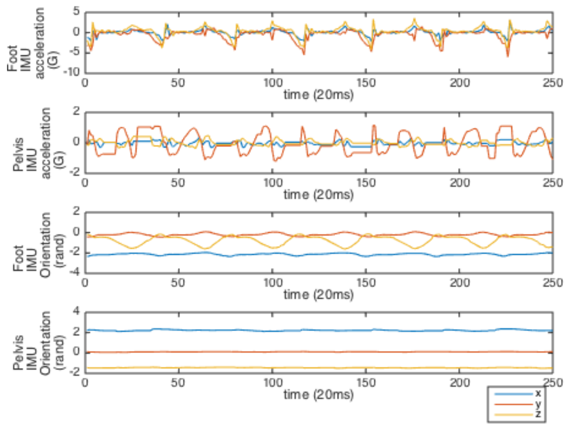

Figure 4. Example of the IMU sensor output.

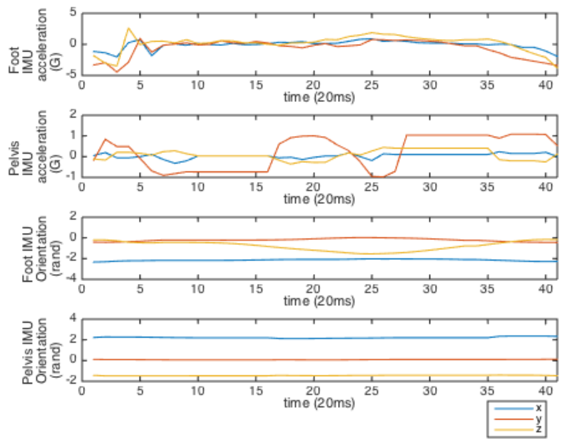

Figure 5. IMU sensor output in one running cycle.

B. Foot and Pelvis Postion Estimation

Fig. 6 shows the positions (in X and Y direction, where X is aligned with the running direction, and Y is aligned with the vertical direction) estimation of the foot and pelvis by the IMU acceleration. The estimated positions were compared with the corresponding marker trajectories in VICON system. The results show that the differences between the positions estimated by the IMU based system and the VICON system were generally within 5 mm.

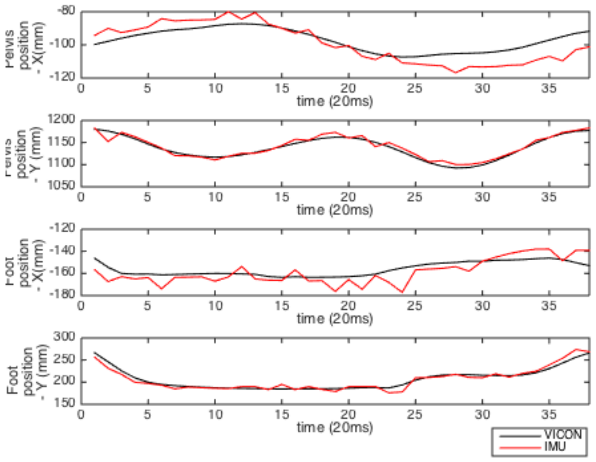

Figure 6. Compare the foot and pelvis position estimated by IMU sensor with the corresponding marker trajectories in VICON system.

C. Lower limb Joint Angle Estimation

Fig. 7 shows 2D (on sagittal plane) joint angles of lower limbs during running estimated by our system in comparison with the data obtained by the motion capture system. The results show that the lower limb joint angles can be estimated accurately. The differences between the estimated 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.

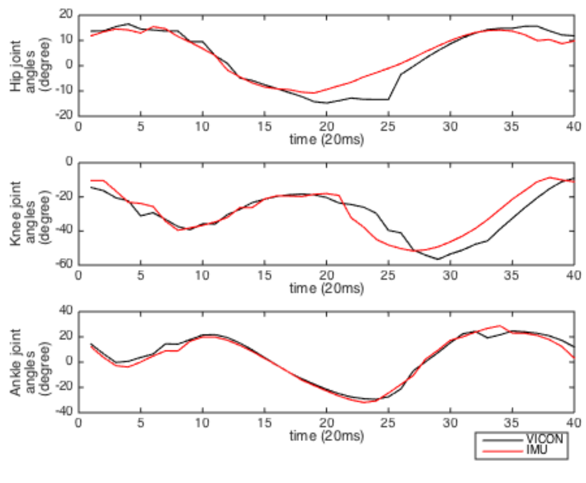

Figure 7. Compare the joint angles estimated by IMU sensor with the VICON system.

IV. Discussions

In this paper, we proposed an IMU sensor based system for real-time biomechanical analysis during Running. The preliminary results based on one vunlenteer show that the lower limb joint angles can be estimated accurately. This allows an real-time feedback implementation for running injury prevention. We have also designed an custermized graphical interface that can monitor the lower limb running posture in real time, as shown in Fig. 8.

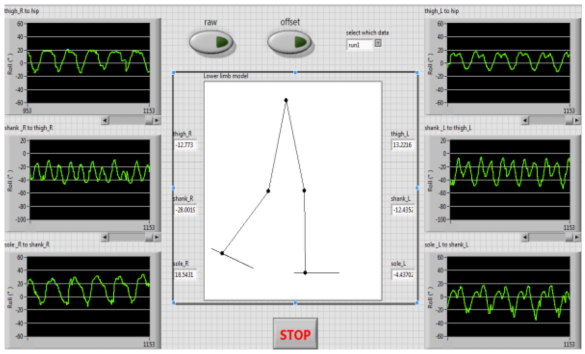

Figure 8. An interface to monitor the running posture in real time.

This study can benefit the future research on conducting complete lower limbs kinematics analysis with minimal and unobtrusive wearable sensors and provide real-time feed back during running exercise. Outdoor experiment will be carried out in the future. Also, the results will be further investigated with more participants involved.

Acknowledgment

This work was supported in part by the Science and Technology Foundation of Shenzhen City, Natural Science Foundation of Guangdong Province, and Start Grand of SZU under Grant number [2016043], [JCYJ20160422145322758], [2016A030310068].

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