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Short communication

Inertial sensor based method for identifying spherical joint center of rotation

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ABSTRACT

The miniaturized wireless inertial measurement unit (IMU) technology and algorithms presented herein promote rapid and accurate predictions of the center-of-rotation (CoR) for ball/spherical joints. The algorithm improves upon existing IMU-based methods by directly utilizing the measured acceleration and angular velocity provided by the IMU to deduce the CoR in lieu of relying on error-prone velocity and position estimates. Results demonstrate that this new method resolves the position of the CoR to within a 3 mm sphere of the true CoR determined by a precision coordinate measuring machine. Such accuracy may render this method attractive for broad use in field, laboratory and clinical settings requiring non-invasive and rapid estimates of joint CoR.

1. Introduction

Functional methods are commonly used in human biomechanics research for determining spherical joint center of rotation (CoR). These techniques rely on video motion capture to track the 3-D position of the body segments on either side of a joint during a prescribed motion. The relative motion of the segments provides the data needed to estimate the joint CoR (Ehrig et al., 2011; Gamage and Lasenby, 2002; Halvorsen et al., 1999; MacWilliams, 2008; Schwartz and Rozumalski, 2005; Siston and Delp, 2006) to within 2.2 mm (MacWilliams, 2008; Siston and Delp, 2006). However, the required motion capture equipment is expensive and data processing is time consuming, thereby constraining use to research laboratory settings. Segment-mounted inertial measurement units (IMUs), which directly measure angular velocity and linear acceleration, may pose an attractive alternative to video-based motion capture for determining joint CoR. The advantages of IMUs derive from their low cost, portability (potential use outside the laboratory and in clinical settings), and high data fidelity.

For instance, the OrthAlign™ knee align system uses a femur-mounted IMU to estimate the location of the CoR of the hip joint for total knee arthroplasty. This system computes CoR position using algorithms similar to those presented in (Gamage and Lasenby, 2002; Halvorsen et al., 1999; MacWilliams, 2008; Schwartz and Rozumalski, 2005; Siston and Delp, 2006) which rely on femur velocity and position estimates (Van der Walt, 2012). These IMU-derived estimates, obtained by successive integrations of the IMU-measured acceleration, are subject to error due to sensor drift (Savage, 2000).

To improve this approach, we propose a new algorithm for estimating the CoR of a spherical joint that avoids any need for (error-prone) velocity and position estimates. This method utilizes solely the acceleration and angular velocity data directly measured by the IMU. The objective of this paper is to introduce this new algorithm and to demonstrate its accuracy via experimental benchmarking.

2. Methods

2.1. IMU hardware and experimental apparatus

This study employs a highly miniaturized wireless IMU which represents the latest in a series of designs developed for sports training and biomechanics studies; refer, for example, to (King et al., 2012, 2010; McGinnis and Perkins, 2012). The IMU is equipped with a low-power Wi-Fi module which wirelessly transmits three axis acceleration and angular velocity data to a host computer. Prior to use, the IMU is calibrated to account for sensor misalignments, scale factors, and biases following (King, 2008). Calibration ensures that the acceleration and angular velocity data are accurately resolved along orthogonal unit vectors defining the IMU frame of reference (yellow arrows in Fig. 1B). For further details, the interested reader may refer to (McGinnis and Perkins, 2012).

The IMU described above is attached to the experimental apparatus illustrated in Fig. 1 which serves as a mechanical approximation of a human hip joint (spherical joint). The joint is formed by a ball bearing (38 mm dia.) that seats in shallow spherical cavities machined into the proximal (black) and distal (white) halves of the joint (Fig. 1A). A pair of tensioned o-rings provides joint pre-loading. The proximal (black, acetabular cup) half is fastened to a table support while the distal (white, femur) half forms a long appendage that may be freely manipulated; refer to Fig. 1B. The IMU (located at yellow frame of reference) is embedded in a calibration jig (black) at the end of the long appendage. The calibration jig serves

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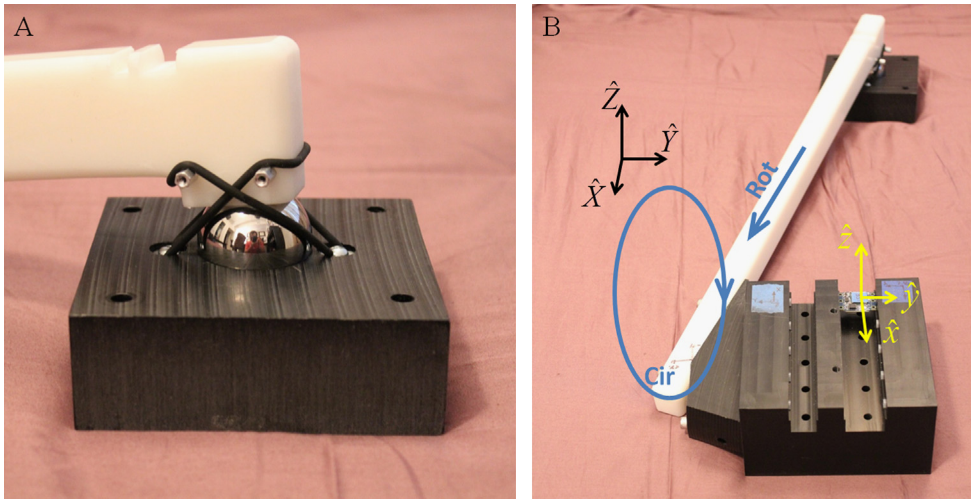

Fig. 1. A mechanical approximation of a human "hip joint" (A) composed of a ball bearing (38 mm dia.) seated between two shallow spherical cavities. One cavity is machined in the proximal side of the joint (black) and the other is machined into the distal side (white). Stretched o-rings provide joint pre-loading. The extension of the distal side (B) supports a machined calibration jig (black) with the embedded wireless IMU. (For interpretation of the references to color in this figure, the reader is referred to the web version of this article.)

as a convenient means to attach the IMU with arbitrarily selected yet measurable position and orientation relative to the center of the bearing (i.e., the joint CoR).

2.2. *Experimental procedure*

The experimental procedure requires recording IMU-measured acceleration and angular velocity data while subjecting the "femur" to two motions: a circumduction (Cir) motion and a rotation (Rot) motion with a pause in between (Fig. 1B). During circumduction, the end of the femur follows the illustrated (near circular) orbit which elicits femoral motion along the surface of a cone with approximately 30° aperture. In human subjects, this would induce modest (and near equal) extension/flexion and ab/adduction of the hip. During rotation, the femur is rotated about its long axis, which would induce (near pure) internal/external rotation of the hip with amplitude ~10° in human subjects. On average, these motions, occurring at a frequency of approximately 1 Hz, induce less than 1 g of acceleration at the accelerometer. According to two orthopedic surgeons consulted for this study, the amplitude of these motions falls well within the standard range of motion (RoM) for the hip, and are characteristic (yet drastically more modest in amplitude) of typical RoM tests performed regularly in clinical settings.

2.3. *Measurement theory*

Following this motion sequence, the IMU-measured acceleration and angular velocity data is transmitted to a host computer for subsequent data analysis. The analysis begins by removing the (constant) 1 g acceleration due to gravity that is detected by the accelerometer in addition to the superimposed acceleration due to motion. To this end, we introduce two frames of reference: an "IMU-frame" denoted by the orthonormal vectors $(\hat{x},\hat{y},\hat{z})$, and an inertial, "lab-frame" denoted by the orthonormal vectors $(\hat{X},\hat{Y},\hat{Z})$; refer to Fig. 1B. The acceleration and angular velocity are measured in the IMU-frame, while gravity is naturally defined in the lab-frame by $-g\hat{Z}$. The acceleration imparted at the accelerometer ($\vec{a}_a$) is recovered from the acceleration measured by the accelerometer ($\vec{a}_m$) per

$$\vec{a}_a = \vec{a}_m - g\hat{Z} \tag{1}$$

which further requires knowledge of the components of $\vec{a}_m$ in the lab-frame. These components are deduced by first computing the direction cosine matrix that defines the orientation of the IMU-frame relative to the lab-frame according to the method presented in (McGinnis and Perkins, 2012). Following these steps allows one to compute $\vec{a}_a$ from Eq. (1) and to also resolve $\vec{a}_a$ in the IMU-frame for use in the following acceleration analysis.

Assuming a rigid femur, the acceleration of the femur-mounted accelerometer $\vec{a}_a$ is related to that of the center of the spherical joint $\vec{a}_c$ through

$$\vec{a}_a = \vec{a}_c + \frac{d\vec{\omega}}{dt} \times \vec{r}_{a/c} + \vec{\omega} \times (\vec{\omega} \times \vec{r}_{a/c}) \tag{2}$$

where $\vec{\omega}$ is the measured femoral angular velocity, $d\vec{\omega}/dt$ is the computed femoral angular acceleration (via numerical differentiation of $\vec{\omega}$), and $\vec{r}_{a/c}$ is the desired but unknown position of the accelerometer relative to the center of the spherical joint. If one assumes that the spherical joint forms the pivot of a spherical pendulum, then $\vec{a}_c = 0$ and Eq. (2) is linear in the remaining unknown $\vec{r}_{a/c}$. Moreover, if one writes Eq. (2) for each of n samples of IMU data, then a solution for $\vec{r}_{a/c}$ can be found using standard least squares.

To demonstrate the accuracy of the proposed algorithm, we present the average (standard deviation) of the three components and length of $\vec{r}_{c/a}$ as determined from 28 trials of IMU data and 14 trials of digital coordinate measuring machine data (CMM - MicroScribe G2X). This CMM, which has positional accuracy/resolution of 0.23/0.13 mm, is used to digitize the location of the center of the accelerometer and 40 points on the surface of the ball bearing (serving as the analog for the femoral head). These 3-D positions are used as input to a surface fitting algorithm that calculates the surface of the bearing and, from this, the true position of the center of the spherical joint relative to the center of the accelerometer $\vec{r}_{c/a}$. Points on the calibration jig (Fig. 1B) are also digitized to define the orientation of the IMU-frame relative to the measurement frame of the CMM. Specifically, 24 points are digitized on the surface of the calibration jig to define the $\hat{x}$–$\hat{y}$ plane, the normal to which defines the $\hat{z}$ axis. Ten points are also digitized along the slot securing the IMU, and the projection of the best-fit line to these points onto the $\hat{x}$–$\hat{y}$ plane, defines $\hat{x}$. Finally, $\hat{y} = \hat{z} \times \hat{x}$. Expressing $\hat{x}, \hat{y}$, and $\hat{z}$ in terms of CMM measurements provides the information necessary to resolve the CMM-measured $\vec{r}_{c/a}$ in the IMU-frame, enabling direct comparison between IMU and CMM estimates.

3. Results and discussion

The experiment and methods above consider an ideal spherical joint defined by an IMU rigidly attached to a rigid femur. This establishes an important limiting case for assessing the accuracy of the new IMU-based method for determining joint CoR. Moreover, this also establishes a direct comparison to a benchmarking study for video-based methods (MacWilliams, 2008) which employs a similar mechanical joint.

Fig. 2 illustrates the IMU data from a representative 60-second trial composed of twice-repeated circumduction and rotation phases. The calibrated angular velocity and acceleration appear in Figs. 2A and B, respectively. The circumduction motions are highlighted by gray boxes and annotated with "Cir" while the rotation motions are highlighted by yellow boxes and annotated with "Rot". Between these motions, the femur is momentarily at rest; observe phases where the angular velocity remains zero and the acceleration remains -1 g. This data is subsequently used to predict the location of the joint CoR following the methods above.

Table 1 summarizes the results from the benchmarking experiment. Reported is the average (standard deviation) of each of the three components and length of the position vector $\vec{r}_{c/a}$ (the center of the spherical joint relative to the center of the accelerometer in the IMU-frame) as independently derived from measurements from the CMM and the IMU. Also reported is the

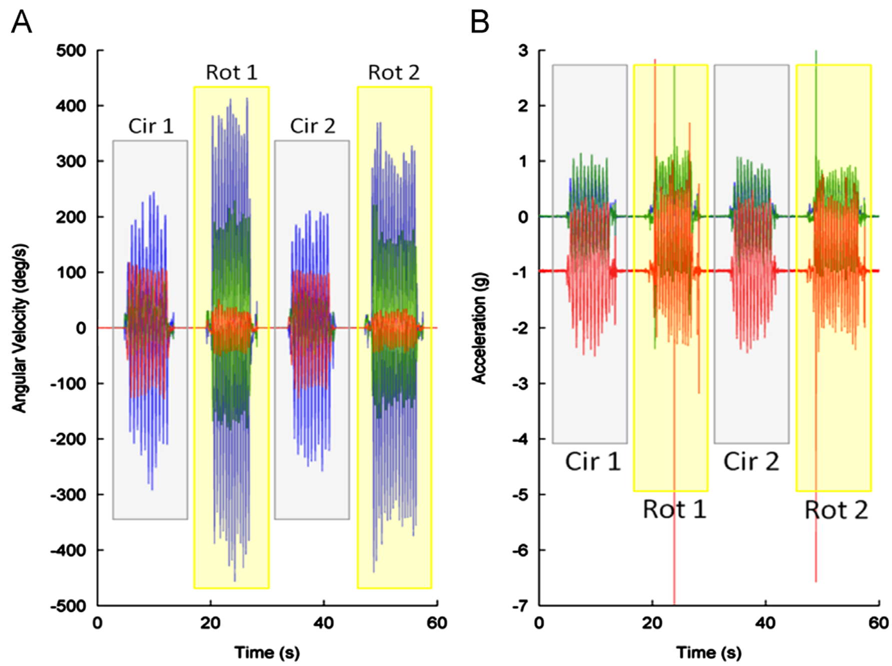

Fig. 2. Three components of angular velocity (A) and three components of acceleration (B) data for an example 60-second trial. Trial consists of two phases of circumduction motion ("Cir" annotation, gray box) followed by two phases of rotation motion ("Rot" annotation, yellow box). Components of angular velocity and acceleration resolved along the IMU-fixed frame are distinguished by the following colors: x axis = blue, y axis = green, z axis = red.

Table 1
Summary of benchmarking experiment, mean (standard deviation) of each component of the joint center position $\vec{r}_{c/a}$ (in mm) for 14 trials of CMM data and 28 trials of IMU data. Third row reports difference in the averages (in mm). Fourth column reports vector sum of the components (in mm).

Method	x (mm)	y (mm)	z (mm)	Length (mm)
CMM	−342.5 (0.4)	288.9 (0.4)	27.9 (0.2)	449.0 (0.3)
IMU	−340.5 (4.4)	290.9 (2.9)	29.1 (1.4)	448.8 (4.8)
(CMM-IMU)	−2.1	−2.0	−1.2	0.2

difference between the average components which further yields an overall error of 3.1 mm (vector sum) for the IMU-derived position relative to the CMM-derived position. The difference in average length of the estimates is also reported, and is only 0.2 mm which suggests one possible source of error stems from identifying $\hat{x}$, $\hat{y}$, and $\hat{z}$ in the CMM measurement frame. However, this overall positional error (3.1 mm) is comparable to the results of (MacWilliams, 2008) where four video capture methods yield average positional errors between 1 and 6 mm. That study employs a similar mechanical spherical joint with rigid marker attachments.

The IMU-derived joint CoR remains within a 3 mm sphere surrounding the true position measured independently from a precision CMM. Importantly, the IMU-method addresses major shortcomings of video-based methods including their high cost, restricted use to motion capture laboratories, long set-up time (attaching and calibrating reflective markers), and long data reduction time. By contrast, the IMU-method requires a single (and inexpensive) segment-mounted IMU, enables use in clinical, field or laboratory settings, and requires only short duration (30–60 s) testing with rapid reporting of results (5 s). These advantages combine to yield a promising non-invasive and accurate tool for estimating joint CoR, provided these methods can effectively be applied to human subjects.

We believe challenges in translating these results to human subjects will likely arise from two simplifications made in this work. First, this experiment considers measurements from an IMU rigidly attached to a rigid femur. In practice, the IMU would be skin mounted and thus the measured acceleration and angular velocity would be polluted by soft tissue motion. However, the relatively slow and gentle motions required for this technique may also induce only modest soft tissue motions. Second, this experiment considers a simulated acetabular cup with a fixed CoR. In practice, a patient's joint center could translate slightly due to motion of the pelvis and/or laxity in the joint. However, we again believe that the circumduction and rotation motions considered herein are unlikely to induce significant soft tissue motion or acceleration of the hip joint/pelvis because they induce modest accelerations (<1 g on average) at the accelerometer. In addition, the ranges of motion are modest relative to standards used in the clinic. However, we recommend that future studies examine both potential limitations through follow-on studies using living or cadaveric subjects.

Conflict of interest statement

The body of this work has been disclosed in a provisional patent application to the USPTO number 61/694,790.

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