IMU-based Joint Angle Measurement Made Practical

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Abstract – This contribution is concerned with hinge joint angle calculation based on inertial measurement data in the context of human motion analysis. Unlike most robotic devices, the human body lacks even surfaces and right angles. Therefore, we focus on methods that avoid to assume a certain mounting orientation, i.e. certain orientations in which the sensor should be mounted with respect to the body segment or to a global coordinate system. After a review of available methods that may cope with this challenge, we present two new methods which use recent results on joint axis and position estimation based on the exploitation of kinematic constraints. Results from an experiment with a transfemoral amputee are provided in which we compare the IMU-based methods to an optical 3d motion capture system. RMS errors are found to be less than 0.6° on the prosthesis and below 4° on the contralateral leg.

1 Introduction

1.1 Inertial Measurement Units

Inertial Measurement Units (IMUs) measure the acceleration, the angular rate and the magnetic field vector in their own local coordinate system (CS). With proper calibration, the axes of the local CS represent an orthonormal base, which is typically well aligned with the outer casing of the sensor.

In addition to the inertial measurement signals mentioned above, some commercially available devices incorporate algorithms that provide online estimates of the sensor’s orientation in a global fixed CS (see e.g. [16]). Typically, this estimation employs strap-down-integration of the angular rates. The drift in the inclination part of the IMU’s orientation is then eliminated using the assumption that the time-averaged acceleration is only gravity [9]. Similarly, the estimation of the IMU’s azimuth (or heading) requires the use of magnetometer measurements. Therefore, it is important to note that long-lasting large accelerations, e.g. in a car or train, and the presence of magnetic disturbances, e.g. ferromagnetic material or mobile phones, may severely limit the accuracy of the orientation estimates (see e.g. [1, 16]). We shall keep these limitations in mind as we analyze the currently available methods for IMU-based knee angle estimation in Section 2.

1.2 Robotic Hinge Joint vs. Human Knee

It has been demonstrated in a large number of publications (see e.g. [3] and references therein) that inertial measurement data can be used to calculate joint angles when at least one IMU is attached to each side of the joint. In this contribution, we will focus on the calculation of hinge joint (or pin joint or knuckle joint) angles, i.e. joints with one rotational degree of freedom, as depicted in Figure 1.

In most robotic and mechanical applications, the sensors can be mounted in such a way that one of the local coordinate axes coincides with the hinge joint axes, see e.g. [10, 3]. In that case, the hinge joint angle can be calculated by integrating the difference of both angular rates around the corresponding coordinate axis. Since even the most precise calibration will yield a non-zero bias, this calculated angle will be subject to drift. However, multiple techniques have been suggested to eliminate this effect using additional information from the accelerometers and/or the magnetometers, see e.g. [3].

Similarly, IMUs can be used to calculate hinge joint angles on the human body, e.g. on the knee joint. Even if, for a moment, we neglect the fact that the knee is not a perfect hinge joint, then there is still a very important difference between the human knee and a robotic hinge joint: It is very difficult to attach IMUs to the leg in such a way that one of the local coordinate axes coincides exactly with the knee joint axis. There have been some attempts, see e.g. [8, 5], but since the human body lacks even surfaces and right angles, the accuracy of such approaches is very limited. In contrast, the body straps that are commonly used to attach IMUs to the leg yield an almost arbitrary orientation of the IMU towards its segment, as illustrated in Figure 2. Nevertheless, the hinge joint angle can be calculated from the inertial measurement data. In
that case, the inertial measurement data from both sensor units must be transformed into a joint coordinate systems [14], i.e. a coordinate systems in which one or two axes coincide with the joint axis and/or the longitudinal axis of the segment. How this transformation can be achieved is discussed in Section 1.3 by reviewing common methods from the literature as well as a rather new approach that exploits the kinematic constraints of the joint. Subsequently, we will analyze how these techniques have been used by different authors to calculate the flexion/extension angle of the knee joint.

1.3 Arbitrary Mounting Orientation and Position

It is a fundamental problem in IMU-based human motion analysis that, due to arbitrary mounting orientation, the IMUs' local coordinate axes are not aligned with any physiologically meaningful axis. However, if the coordinates of the joint axis in both local sensor frames are known, then the measured accelerations, angular rates, and magnetic field vectors of each sensor can be transformed into a CS in which one axis is aligned with the joint axis. In this new coordinate system, flexion/extension corresponds to just one component of the angular rate vectors. In a similar way, one coordinate axis can be aligned with the longitudinal axis of the segment. In result, the inclination of that axis equals the inclination of that segment.

First, we shall note that in some publications this problem is ignored completely by assuming that the IMUs can be mounted precisely in a predefined orientation towards the joint, see e.g. [8, 5]. As can also be seen in the figures therein, this is a rather rough approximation. In the more realistic case of arbitrary mounting orientation, it is a common approach to identifying the joint axis coordinates in both local frames through calibration postures and/or calibration movements. Some authors, e.g. [6, 13], make the subject stand with vertical, straight legs for a few seconds and use the acceleration measured during that time interval to align a local coordinate axis with the longitudinal axis of the segments. In addition, sitting calibration posture are used in [13]. Besides static postures, predefined calibration motions can be used to identify the coordinates of physically meaningful axes in the upper and lower sensor CS. Examples can be found in [6, 7, 11].

However, it is important to note that, in both calibration postures and calibration motions, the accuracy is limited by the precision with which the subject can perform the postures or motions. In case this limitation is undesirable, a third technique can be used, that has been developed recently. In [12], the inertial measurement data from almost arbitrary motions has been used to identify the coordinates of the joint axis. This is done by exploiting the kinematic constraints of the joint. In addition to identifying the knee axis, this approach also allows for precise estimation of the joint position in both local CSs, see Figure 2 for an illustration. That piece of information is required in some joint angle algorithms, see e.g. [8], and is difficult to be determined by manual measurements. Furthermore, it has been demonstrated in [15] that the joint position vectors can be used to improve the accuracy of body segment orientation estimates.

The reviewed methods of joint axis and position identification make a major contribution to the quality of IMU-based joint angle measurements, since the alternative, i.e. manual measurements, can hardly yield accurate estimates, as demonstrated e.g. in [12, 8]. Therefore, most of the methods reviewed in the next
section employ such techniques.

2 Brief Review of IMU-based Knee Angle Estimation

Many algorithms and techniques have been suggested for IMU-based knee angle estimation. Despite the variety of approaches, the vast majority of authors defines the flexion/extension angle of the knee joint as the angle between the upper and lower leg along the main axis of relative motion, i.e. the knee joint axis [4, 6, 8, 13]. In other words, the projections of the upper and lower leg into the joint plane, to which the joint axis is normal, confine this angle.

However, we shall note that speaking of the knee as a hinge joint is an approximation. Although flexion and extension is the major degree of freedom, a biological joint such as the knee is not perfectly constrained to rotation around one axis. This is often addressed by additionally considering abduction/adduction and internal/external rotation, which leads to a three-dimensional knee joint angle, as e.g. in [6, 2, 5]. However, abduction/adduction and internal/external rotation angles hardly ever exceed a range of ±10° (cf. [6]). Therefore, these additional degrees of freedom are neglected in some publications, e.g. in [4, 8, 13].

The most simple approaches in the literature simply assume that the IMUs are attached such that one of the local coordinate axes is aligned with the joint axis. As mentioned above, integrating the difference of the upper and lower sensor’s angular rates around that axis will yield a drifting flexion/extension angle. In [5], this drift was removed using a high-pass filter. In another publication with the same mounting assumption, it was demonstrated that the joint angle can also be estimated from the measured accelerations if the position of the joint in both local CSs is known [8]. Thereby, a root mean square error (RMSE) of less than 4° with respect to an optical reference system was achieved. Although both techniques may seem restricted to a special sensor mounting, they are just as helpful in the case of arbitrary mounting orientation, as long as the joint axis coordinates are known.

A fundamentally different approach is found in [13]. After identifying the segment’s longitudinal axis coordinates, the authors calculate the thigh’s and the shank’s inclination and approximate the flexion/extension angle by the difference of these inclinations. Thereby, they achieve a RMSE of approximately 7° with respect to an optical reference system. However, their method is bound to the assumption that the knee axis remains horizontal during the entire motion. While that might be an acceptable approximation for most walking and running situations, this assumption does not hold during quick direction changes and for a number of other motions like skating, hurdles or martial arts. In [4], the aforementioned method has been advanced. Instead of assuming a horizontal knee axis, the authors model the knee as a pure hinge joint and exploit its kinematic constraints using an extended Kalman filter. Thereby, they are able to calculate flexion/extension angles in well accordance with an optical reference system, both at the speed of running (8km/h, RMSE<4°) and walking (3km/h, RMSE<1°). Approximately the same precision for walking is achieved in [6]. Here, however, the complete orientation of each IMU with respect to its global reference CS is calculated using a fusion algorithm that combines gyroscope and accelerometer measurements. Just as before, both of these two last approaches assume that the knee joint axis has been identified using one of the methods in Section 1.3.

Which of the reviewed methods is most suitable for a specific application depends also on the available sensor information. In [13, 4, 6], the orientations of the thigh and shank are used to calculate the flexion/extension angle. This is straightforward if reliable sensor orientation estimates are available, and if the joint axis coordinates in both local CSs are known. However, knowing the joint axis allows to reduce the problem to one dimension immediately. Therefore, it might be advantageous to use one (or a combination) of the methods in [8, 5]—especially if reliable orientation estimates are not immediately available.

3 New Methods exploiting kinematic constraints

In this section, we introduce two methods for IMU-based knee joint angle estimation both of which can handle arbitrary mounting orientation and position. They use parts of the methods reviewed above but avoid mounting assumptions as well as manual measurements by exploiting kinematic constraints instead. Precisely, the first two steps of both methods are the following: The subject is instructed to perform ten seconds of upper and lower leg circling with a few arbitrary changes in direction, amplitude and frequency of the circling motion (see e.g. [17]). Subsequently, the least-squares method in [12] is applied to the measured accelerations and angular rates to identify the knee joint axis in both local sensor CSs. As outlined above,
Figure 3: Two algorithms for IMU-based knee angle calculation are evaluated experimentally. \textit{Left:} Sensor orientation estimates are used to calculate the orientational difference (i.e. the joint angle) around a given axis. \textit{Right:} The problem is reduced to one dimension immediately by integrating the difference of the angular rates around the joint axis. Then, an acceleration-based joint angle estimate is used to remove drift.

this information is crucial to precise joint angle calculation. It is used to transform the measurement data into joint-related CSs, in which the z-axes coincide with the joint axis. In addition, the measurement data from the circling movement is used to calculate the joint position vectors in both local sensor CSs. This information, however, will only be used by the second method.

The first method assumes that each IMU provides highly accurate estimates of its orientation with respect to a fixed reference CS. This orientation information is used to calculate the knee joint angle. To this end, the same method as in [6] is used, but only the flexion/extension angle is calculated. Therefore, is is sufficient to know one common axis (here: the knee joint axis) in both local CS rather than two axes, as in [6].

The second method reduces the problem to one dimension from the very start by integrating both angular rates only around the joint axis. As the difference of both integrals, a joint angle estimate is obtained that is highly accurate up to slow drift. In a second step, a noisy but driftless joint angle estimate is calculated from the measured accelerations and the joint position vectors, as explained in [8]. Finally, these two estimates are combined by a Kalman filter to obtain a highly accurate, driftless flexion/extension angle.

4 Experimental Results and Discussion

The two methods that were introduced in Section 3 are now evaluated in repeated gait experiments with a transfemoral amputee wearing a leg prosthesis. We use elastic body straps to attach one IMU (xsens MTw [16]) each to the upper and lower leg of both the prosthesis and the contralateral leg, as depicted in Figure 2. We neither restrict the mounting to certain locations or orientations, nor do we measure these quantities. Instead, the two methods that were introduced in Section 3 are used to calculate the flexion/extension angles of both legs, while the subject walks about ten meters on a straight line within the range of an optical gait analysis system and far away from potential magnetic disturbances.

The resulting joint angle traces are provided in Figure 4. With respect to the optical system, both IMU-based methods achieve a root-mean-square deviation of less than 0.6° on the prosthesis side and less than 4° on the contralateral side. One possible reason for the larger deviations on the contralateral side might be that the human knee is less close to a perfect hinge joint. This fact, however, was taken care of by calculating only the flexion/extension component of the joint angle from both the inertial and the optical data. A much more reasonable explanation is that, unlike previous publications, we decided \textit{not} to attach the optical markers to the IMUs rigidly (see e.g. [6, 11]). Instead, they were placed in standard positions for optical gait analysis, see Figure 2. Therefore, the effects of skin and muscle motions become apparent in the angle traces of the contralateral side. The deviations between optical and IMU-based angles are largest during pre-swing and heel strike, i.e. when the leg is accelerated and decelerated. This effect is not observed on the prosthesis side, where optical markers and IMUs are connected by rigid bodies (i.e. the prosthesis segments).
Figure 4: Comparison of the two IMU-based knee angle measurements ($\alpha_{\text{IMU-or.}}$ and $\alpha_{\text{IMU-ax.}}$) with the result ($\alpha_{\text{opt.}}$) of an optical gait analysis system. Only flexion/extension is considered. On the prosthesis side, there is no relevant deviation ($e_{\text{pr}}, \hat{e}_{\text{pr}} < 0.6^\circ$). But on the contralateral side, skin and muscle motions effects, which are strongest during pre-swing and heel-strike, lead to RMS errors $e_{\text{cl}}$ and $\hat{e}_{\text{cl}}$ of almost 4°.

5 Conclusion

Two new IMU-based joint angle measurement techniques were proposed that exploit kinematic constraints of the joint. Both techniques use elements of previously suggested methods in a new combination that can handle arbitrary mounting orientation and position – a fundamental challenge in IMU-based gait analysis. Especially when working with stroke patient and amputees, we believe that these methods are more practical than methods that require the patient to perform a precise calibration movement or methods that require to attach the sensors in specific positions or orientations.

The first method that was introduced requires the sensor to provide precise orientation estimates and thus also relies on magnetometers. Besides this method, an equally accurate method was suggested that uses only accelerometers and gyroscopes. Since magnetometers are excluded, this method can be used indoors and in the proximity of magnetic disturbances. Furthermore, it is straightforward to implement both algorithms online. This also applies to the least-squares technique that is used in both methods to estimate the local joint axis (and position) coordinates. Therefore and since they supersede manual measurements and precise calibration movements, these new methods open the door to a plug-and-play gait analysis, in which one simply attaches the IMUs, moves the joints a few seconds, and then receives joint angle measurements in real-time. This will be subject of further research.

References


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