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Deriving kinematic quantities from accelerometer readings for assessment of functional upper limb motions

Abstract: Wearable accelerometers are lightweight, affordable, and allow for even smaller form factors than 9D inertial measurement units. They are therefore a promising tool for assessing the quality of movement of patients during daily life activities. While generic signal features such as signal power and frequency content are widely used, the derivation of kinematic (angular and spatial) quantities remains a challenge. We consider a chain of body segments, such as the arm, equipped with 3D accelerometers and propose a method for calculation of the inclination and relative height of the distal segment. For validation of the method against an optical motion capture system, we consider a setup with accelerometers on the forearm and the upper arm of a subject, who performs a sequence of drinking motions and pick-and-place motions. We obtain a root-mean-square deviation of about 2.5 cm for the wrist height relative to the shoulder and about 6° for the inclination angles of the forearm. We conclude that the proposed method yields measurements of kinematic quantities that are accurate enough for classification of functional versus non-functional motions or well-performed motions versus incomplete motions.

Keywords: Inertial motion capture, accelerometer, rehabilitation, biomechanics, limb motion analysis, activities of daily living.

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1 Introduction

In stroke rehabilitation and similar fields it is desirable to assess the quality of functional lower and upper limb motions, for example in activities of daily living. Various sensor technologies have been proposed for this purpose. Marker-based optical motion capture is considered a gold standard for accuracy, but it is confined to expensive laboratory environments. Since they are more affordable and suitable for ambulatory use, inertial sensors, which combine accelerometers, gyroscopes and magnetometers, have been proposed as an alternative [2, 7]. However, as we aim to demonstrate, a similarly good measurement of kinematic quantities can be obtained by using only accelerometers if heading information is not required and if the considered motions are not very fast.

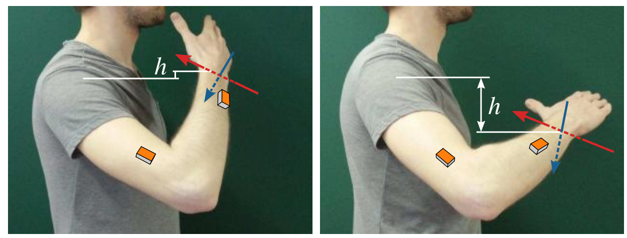

Figure 1: Functional and non-functional drinking pose. The depicted poses differ by the height of the wrist with respect to the shoulder (h) and by the forearm orientation, which is well described by a thumb vector (red) and a palm vector (blue).

Accelerometers have previously been used to analyze daily-life upper limb motions, see [4] for a recent review. However, analysis is usually based on activity measures like duration of use [8], movement counts [3], activity asymmetry indices [6], or energy and entropy measures [1]. While such generic parameters describe general characteristics of a motion, we believe that calculating kinematic quantities (such as vertical positions, orientations and velocities) yields much deeper insight and allows for a more detailed, accurate and robust classification of functional motions.

In the current contribution, we propose a method for assessing static poses and dynamic motions of the upper limb by estimating three kinematic quantities. The first quantity is the height h of the wrist (i.e. the distal end of the forearm) relative to the shoulder (i.e. the rotation center of the glenohumeral joint). The second and third quantities are the inclinations of the two wrist axes shown in Fig. 1, which correspond to the directions of the extended thumb and the normal vector of the palm in a neutral wrist position. For

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sake of simplicity, we name these signals the "thumb-up" angle θ_t and "palm-up" angle θ_p.[1]

As Fig. 1 indicates, these three quantities can be used to differentiate between a proper, functional motion and an improper, non-functional motion. Therefore, a method that determines these three quantities accurately facilitates the assessment of upper limb poses and motions.

2 Method

To calculate the wrist height and inclination signals in realtime, we attach two triaxial accelerometers to the subject's arm, one on the upper arm and one on the forearm close to the wrist. We assume that the coordinates of the segment axes in the local sensor frames, i.e. $(\mathbf{i}_1, \mathbf{j}_1, \mathbf{k}_1)$ and $(\mathbf{i}_2, \mathbf{j}_2, \mathbf{k}_2)$ as shown in Fig. 2, either coincide roughly with the local sensor axes or are known by attachment. Furthermore, we use the fact that the ratio $r = l_1/l_2$ of the upper arm length and the forearm length is 1.22 on average and varies by less than 7% between individuals [5].[2]

We measure the upper arm acceleration $\mathbf{a}_1(t)$ and the forearm acceleration $\mathbf{a}_2(t)$ at a sampling rate $f_s \geq 10$ Hz. These accelerations are filtered using a digital lowpass filter (2nd-order Butterworth) with a cutoff frequency of 1 Hz and the resulting vectors are normalized. This yields an estimate of the vertical unit vector in the sensor's local coordinates. We denote those signals by $\hat{\mathbf{a}}_1(t)$ and $\hat{\mathbf{a}}_2(t)$, respectively.

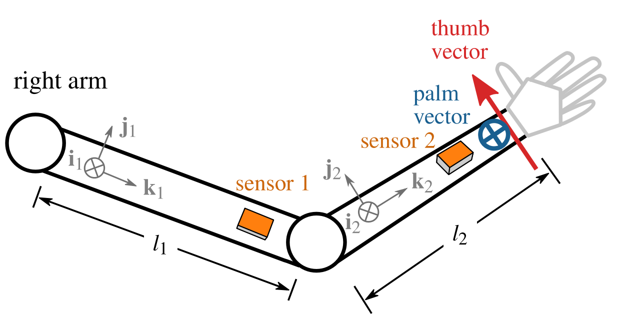

Figure 2: Kinematic model of upper arm with coordinate system $(\mathbf{i}_1, \mathbf{j}_1, \mathbf{k}_1)$ and length l_1 and forearm with coordinate system $(\mathbf{i}_2, \mathbf{j}_2, \mathbf{k}_2)$ and length l_2. Triaxial accelerometers are attached to both segments with known orientation.

1 Note that we neither track the hand motion, nor do we assume certain wrist or finger joint angles. The names are simply chosen to verbalize the orientation of the forearm in an illustrative way.
2 Note that this assumption is only made to avoid the need for measuring l_1 and l_2 in each individual, and that the method is not restricted to humans with a certain value of r.

2.1 Relative wrist height

For each sample instant, we calculate the height of the wrist relative to the shoulder as

$$h(t) = \frac{r}{r+1}\mathbf{k}_1 \cdot \hat{\mathbf{a}}_1(t) + \frac{1}{r+1}\mathbf{k}_2 \cdot \hat{\mathbf{a}}_2(t). \tag{1}$$

Note that this yields the vertical wrist position in relative dimensions, where $h = 0$ refers to the wrist being on the same height as the shoulder and $h = -1$ refers to the wrist being one arm length below the shoulder, i.e. the upper arm and forearm are pointing straight down.

2.2 Wrist inclination

For each sample instant, we calculate the inclination of the axes $\mathbf{i}_2$ and $\mathbf{j}_2$ of the forearm which gives us the "thumb-up" angle

$$\theta_t(t) = \frac{\pi}{2} - \arccos\big(\mathbf{j}_2 \cdot \hat{\mathbf{a}}_2(t)\big) \tag{2}$$

and the "palm-up" angle

$$\theta_p(t) = \frac{\pi}{2} - \arccos\big(\mathbf{i}_2 \cdot \hat{\mathbf{a}}_2(t)\big). \tag{3}$$

Note that the angles are defined so that an angle of 90° means that thumb/palm are pointing up.

2.3 Quaternions for 3D visualization

In order to visualize the pose information that the accelerometers reveal, we calculate quaternions that represent the inclinations of upper arm and forearm. As the heading of the segments cannot be determined from accelerometer readings, we visualize upper arm and forearm separately, each from a view that makes the segment's longitudinal axis align with the x-axis of the visualization space, cf. Fig. 3.

For the right upper arm, we obtain the described quaternions by first expressing the vertical vectors in the coordinate systems of the upper arm (as opposed to the sensor coordinate system), i.e.

$$\hat{\mathbf{a}}_u = [\hat{\mathbf{a}}_1 \cdot \mathbf{i}_1 \quad \hat{\mathbf{a}}_1 \cdot \mathbf{j}_1 \quad \hat{\mathbf{a}}_1 \cdot \mathbf{k}_1]^T. \tag{4}$$

We then define vertical and horizontal reference vectors as $\mathbf{r}_v = [0 \ \ 0 \ \ 1]^T$ and $\mathbf{r}_h = [1 \ \ 0 \ \ 0]^T$ and calculate the quaternion that represents the inclination by

$$\mathbf{q}_{\text{incl}} = \begin{bmatrix} \cos\left(\frac{\alpha}{2}\right) \\ \sin\left(\frac{\alpha}{2}\right)\mathbf{u} \end{bmatrix}^{\mathrm{T}}, \tag{5}$$

with

$$\alpha = \arccos\,(\hat{\mathbf{a}}_{\mathrm{u}} \cdot \mathbf{r}_{\mathrm{v}}), \tag{6}$$

$$\mathbf{u} = \frac{\hat{\mathbf{a}}_{\mathrm{u}} \times \mathbf{r}_{\mathrm{v}}}{\|\hat{\mathbf{a}}_{\mathrm{u}} \times \mathbf{r}_{\mathrm{v}}\|}. \tag{7}$$

The resulting quaternion will always have a rotation axis $\mathbf{u}$ that lies in the horizontal plane and can therefore be regarded as a pure inclination quaternion. To assure the desired heading alignment with the visualization space, we transform the longitudinal axis into the fixed frame[3]

$$[\mathbf{k}_1]_{\mathrm{ff}} = \mathbf{q}_{\text{incl}} \otimes \mathbf{k}_1 \otimes \mathbf{q}_{\text{incl}}^{-1}, \tag{8}$$

project it into the horizontal plane

$$[\mathbf{k}_1]_{\mathrm{h}} = [\mathbf{k}_1]_{\mathrm{ff}} - ([\mathbf{k}_1]_{\mathrm{ff}} \cdot \mathbf{r}_{\mathrm{v}})\,\mathbf{r}_{\mathrm{v}} \tag{9}$$

and calculate a heading quaternion

$$\mathbf{q}_{\text{heading}} = \begin{bmatrix} \cos\left(\frac{\beta}{2}\right) \\ \sin\left(\frac{\beta}{2}\right)\mathbf{v} \end{bmatrix}^{\mathrm{T}}, \tag{10}$$

with

$$\beta = \arccos\left(\frac{[\mathbf{k}_1]_{\mathrm{h}}}{\|[\mathbf{k}_1]_{\mathrm{h}}\|} \cdot \mathbf{r}_{\mathrm{h}}\right), \tag{11}$$

$$\mathbf{v} = \frac{[\mathbf{k}_1]_{\mathrm{h}} \times \mathbf{r}_{\mathrm{h}}}{\|[\mathbf{k}_1]_{\mathrm{h}} \times \mathbf{r}_{\mathrm{h}}\|}. \tag{12}$$

The product $\mathbf{q} = \mathbf{q}_{\text{heading}} \otimes \mathbf{q}_{\text{incl}}$ is used to visualize the right upper arm orientation. The quaternions for the left upper arm and both forearm orientations are obtained analogously. However, in order to obtain correct quaternions for the left arm, the sign of $\mathbf{j}_1/\mathbf{j}_2$ has to be changed.

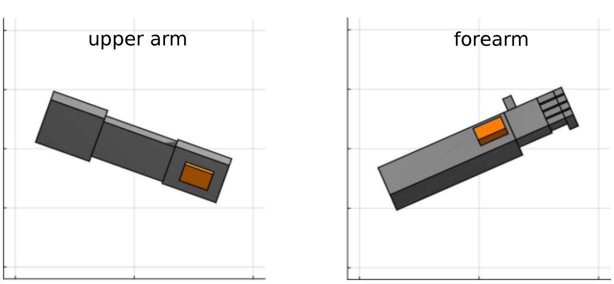

Figure 3: 3D visualization of drinking pose with hand in front of mouth. $h = -0.5$ cm, $\theta_{\mathrm{t}} = 41°, \theta_{\mathrm{p}} = -40°$.

3 We use $\otimes$ to denote quaternion multiplication. When three-dimensional vectors $\mathbf{v} \in \mathbb{R}^3$ are multiplied with quaternions, we regard them as their corresponding pure quaternions $[0\ \mathbf{v}^{\mathrm{T}}]^{\mathrm{T}}$.

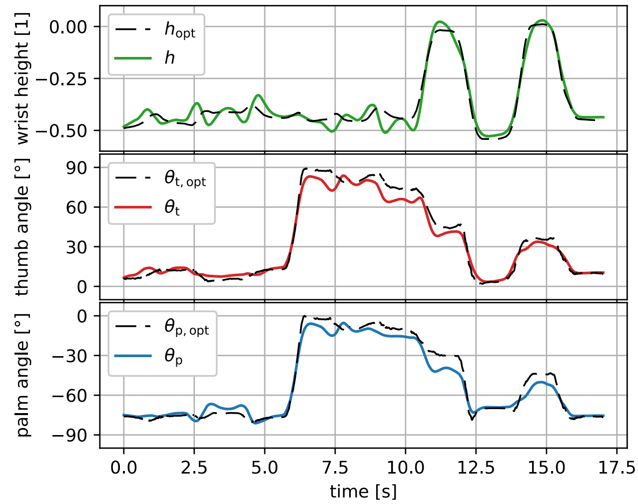

Figure 4: Wrist height and inclination over time for a sequence of pick-and-place and drinking motions with reference quantities derived from optical motion capture.

3 Experimental validation

We validate the accuracy of the proposed method with respect to a marker-based optical motion capture system (Vicon Motion Systems Ltd. UK). Optical markers are attached to the subject on anatomical landmarks, and two inertial sensors (XSens MTw, Xsens Technologies B.V., Netherlands) are placed on upper arm and forearm as shown in Fig. 1.

The subject performs twelve trials consisting of pick-and-place and drinking motions with two different velocities. We employ the method described in Section 2 to calculate the wrist height and inclination from the accelerations recorded by the IMU at $f_{\mathrm{s}} = 75$ Hz. In order to obtain a reference, we use the optical markers to determine the true length of the upper arm ($l_1 = 32.3$ cm) and forearm ($l_2 = 25.9$ cm) and then calculate the true values of the three kinematic quantities.

We timeshift the angles to compensate for the lowpass filter delay. The resulting height and inclinations for one trial are shown in Fig. 4. The root-mean-square error (RMSE) between the accelerometer-based and optical quantities is calculated for all trials. On average, an RMSE of 0.0387 (2.25 cm) for the wrist height, 4.86° for the “thumb-up” inclination and 5.89° for the “palm-up” inclination is obtained.

To investigate the influence of the lowpass filter cutoff frequency, we calculate the RMSE for different cutoff frequencies, as shown in Fig. 5.

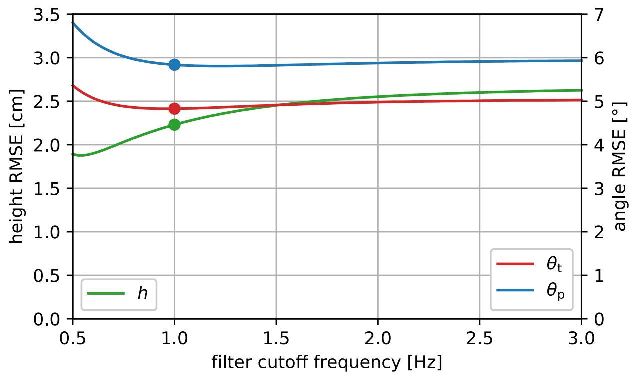

Figure 5: RMSE for different lowpass filter cutoff frequencies. Note that the influence on the quality of the results is small.

4 Discussion

Although the subject's actual humerus-ulna ratio of 1.25 differs from the average value that the method assumes, the wrist height and inclination are measured accurately. The results further indicate that the method has a low sensitivity with respect to the lowpass filter cutoff frequency. The RMSE of the height and inclination angles are robustly in the range of 2.5 cm and 6°, respectively. This is the same level of accuracy as in state-of-the-art inertial motion capture [2]. It seems reasonable to assume that the difference between a functional and a non-functional or incomplete motion is much larger than the obtained deviations.

5 Conclusions

We introduced a method for calculation of kinematic quantities that allow assessment of upper limb motions in activities of daily living. Using optical motion capture as reference, we evaluated the accuracy of the proposed method. We conclude that the measurements are accurate enough for classification of functional versus non-functional motions or well-performed motions versus incomplete motions. Since numbers are less intuitively interpretable than graphical representations, we proposed an effective method for visualization of the measured kinematic quantities. Since all proposed methods are realtime in the sense that they provide the measured quantities with a time delay of less than 0.25 s, the visualization might well be used for functional biofeedback. Future research will be dedicated to deriving specific features from the calculated signals and applying the algorithm to patient data in order to improve classification results.

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Author's Statement

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References

[1] B. Knorr, R. Hughes, D. Sherrill, J. Stein, M. Akay, and P. Bonato. Quantitative measures of functional upper limb movement in persons after stroke. In Neural Engineering, 2005. Conference Proceedings. 2nd International IEEE EMBS Conference on, pages 252–255. IEEE, 2005.

[2] P. Müller, M. A. Begin, T. Schauer, and T. Seel. Alignment free, self-calibrating elbow angles measurement using inertial sensors. IEEE Journal of Biomedical and Health Informatics, 21(2):312–319, 2017.

[3] E. Narai, H. Hagino, T. Komatsu, and F. Togo. Accelerometer-based monitoring of upper limb movement in older adults with acute and subacute stroke. Journal of Geriatric Physical Therapy, 39(4):171–177, 2016.

[4] M. Noorkõiv, H. Rodgers, and C. I. Price. Accelerometer measurement of upper extremity movement after stroke: a systematic review of clinical studies. Journal of neuroengineering and rehabilitation, 11(1):144, 2014.

[5] A. Pietak, S. Ma, C. W. Beck, and M. D. Stringer. Fundamental ratios and logarithmic periodicity in human limb bones. Journal of anatomy, 222(5):526–537, 2013.

[6] M. Rabuffetti, P. Meriggi, C. Pagliari, P. Bartolomeo, and M. Ferrarin. Differential actigraphy for monitoring asymmetry in upper limb motor activities. Physiological Measurement, 37(10):1798, 2016.

[7] T. Seel, D. Graurock, and T. Schauer. Realtime assessment of foot orientation by accelerometers and gyroscopes. Current Directions in Biomedical Engineering, 1(1):466–469, 2015.

[8] G. Uswatte, W. H. Miltner, B. Foo, M. Varma, S. Moran, and E. Taub. Objective measurement of functional upper-extremity movement using accelerometer recordings transformed with a threshold filter. Stroke, 31(3):662–667, 2000.