ELSEVIER

Journal of Biomechanics 40 (2007) 78–85

JOURNAL
OF
BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Ambulatory measurement of arm orientation

H.J. Luinge[a], P.H. Veltink[b,*], C.T.M. Baten[c]

[a]*Xsens Technologies b.v., Enschede, The Netherlands*
[b]*Signals and Systems Group, Department of Electrical Eng., Institute for Biomedical Engineering, University of Twente, P.O. Box 217, 7500 AE, Enschede, The Netherlands*
[c]*Roessingh Research and Development, Enschede, The Netherlands*

Accepted 17 November 2005

Abstract

In order to evaluate the impact of neuromuscular disorders affecting the upper extremities, the functional use of the arm need to be evaluated during daily activities. A system suitable for measuring arm kinematics should be ambulatory and not interfere with activities of daily living. A measurement system based on miniature accelerometers and gyroscopes is adequate because the sensors are small and do not suffer from line of sight problems. A disadvantage of such sensors is the cumulative drift around the vertical and the problems with aligning the sensor with the segment.

A method that uses constraints in the elbow to measure the orientation of the lower arm with respect to the upper arm is described. This requires a calibration method to determine the exact orientation of each of the sensors with respect to the segment. Some preliminary measurements were analyzed and they indicated a strong reduction in orientation error around the vertical. It seemed that the accuracy of the method is limited by the accuracy of the sensor to segment calibration.

Keywords: Upper-extremities; Elbow; Kinematics; Inertial sensors; Gyroscope; Accelerometer

1. Introduction

Measurement of upper extremity kinematics is required in the field of rehabilitation, medicine and ergonomics. For applications which require that these measurements are performed outside the laboratory, body mounted sensors like accelerometers and gyroscopes can be used. Micromachined accelerometers and gyroscopes are suitable for measuring arm movements, since they are sufficiently small to be attached to the upper and forearm without interfering with the subjects' movements.

The assessment of neurological disorders is often conducted by measuring arm movements. For example, Beer et al. (2000) measured the path of the hand in pointing tasks for quantifying hemiparesis, Goldvasser et al. (2001) for quantifying ataxia and Topka et al. (1998) for quantifying dyskinesia. Symptoms of Parkinson's disease were measured using accelerometers (Dunnewold et al., 1997; Hoff and Hilten, 1999). Uswatte et al. (2000) and Bernmark and Wiktorin (2002) used an accelerometer attached to the arm in order to obtain a measure of arm function during daily life. In ergonomics, the measurement of arm movement is important for load estimation.

Suitable sensors for ambulatory measurement of human body orientation are accelerometers, gyroscopes, magnetometers and goniometers. Each of these sensors have different characteristics, advantages and disadvantages. Micromachined accelerometers are small, relatively cheap and have a low energy consumption. They measure acceleration and gravity and can be used as an inclinometer for movements in which the acceleration can be neglected with respect to the gravity (Luinge and

---

*Corresponding author. Tel.: +31 53 489 2760; fax: +31 53 489 2287.
E-mail address: p.h.veltink@utwente.nl (P.H. Veltink).

0021-9290/$ - see front matter © 2005 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2005.11.011

Veltink, 2004; Willemsen et al., 1990). Gyroscopes measure angular velocity, which can be used to estimate a change in orientation. The drawback of gyroscopes is that the estimation of orientation change is prone to integration drift. Magnetometers are used to measure the local earth magnetic field vector. This provides additional information about orientation. However, the fixed magnetic field, and thus the derived orientation, is disturbed in the vicinity of ferromagnetic metals and by electronic equipment generating magnetic fields. Another method is to use goniometers, which measure the angle between two joints but not the inclination with respect to gravity. Because goniometers cross a joint, they need to be exactly aligned with the joint rotation center. Also cross talk is a problem in goniometers (Hansson et al., 1996).

Several combinations of the sensors described above have been proposed in order to overcome the drawbacks of the separate sensors. Kemp et al. (1998) combined a triaxial accelerometer and a triaxial magnetometer to measure an orientation. A triaxial gyroscope and triaxial accelerometer were applied by Foxlin et al. (1998) and Luinge and Veltink (2005) who developed a Kalman filter for measuring orientation. The change in orientation obtained using gyroscopes was fused with the inclination measured by the accelerometers, yielding an inclination estimate that was sufficiently accurate even in the presence of accelerations. However, the error in rotation around the vertical could not significantly be reduced. Bachman (2000) and Zhu and Zhou (2004) used magnetometers in addition to gyroscopes and accelerometers to overcome this problem. Heading errors due to magnetic field disturbance can be effectively rejected by adequate model-based sensor fusion (Roetenberg et al., 2005).

Another method to obtain kinematics between 2 body segments is to estimate the orientations of each segment using a multiple sensor system and use anatomical constraints to link the different segments. This method was applied for measurement of trunk position with respect to the pelvis, required for the ambulatory measurement of low back load (Baten et al., 2002). Using such a method to obtain the orientations of the arm segments may be suitable since the constraints of the elbow angles are relatively well described.

Aim of this study is to derive a method for a drift-free estimate of the orientation of the 2 arm segments using inertial sensors and anatomical elbow constraints. These orientations could be the basis for assessing relative positions of the hand with respect to the shoulder. Initially, the sensor orientation is related to the segment orientation. Subsequently, the orientation between segments is estimated. The orientation estimate is based on the algorithm described in (Luinge and Veltink, 2005) for orientation measurement using gyroscopes and accelerometers as well as the assumption that the elbow joint does not permit adduction.

2. Methods

The orientation of the upperarm with respect to the forearm was measured using an inertial measurement unit (IMU) consisting of 3 gyroscopes and 3 accelerometers. The measurement procedure consisted of 2 stages. First of all, a sensor to segment calibration was conducted in order to find the orientation of the IMU with respect to the segment to which it is attached. Secondly, the orientations of the upperarm and forearm were obtained using angular velocity and accelerometer signals described in the segment coordinate frame. The Kalman filter described by Luinge and Veltink (2005) was used to obtain an orientation estimate suffering from a slowly increasing heading error. The heading was defined as that part of the orientation that describes the rotation around the vertical. Thirdly, heading error between the 2 segments was minimized using the knowledge that abduction/adduction of the elbow joint is constrained.

2.1. IMU-segment orientation calibration

An IMU attached to a rigid body segment measures signals that are expressed in the sensor coordinate frame. If these signals are to be expressed in the coordinate frame of a body segment, the orientation of the IMU with respect to the segment is required. This orientation was obtained by recording the IMU signals while the subject performed several predefined movements.

The coordinate systems of the upperarm and the forearm are defined according to van der Helm and Pronk (1995) as shown in Fig. 1. The forearm IMU was placed on the dorsal side of the forearm, near the wrist. The upper arm IMU was placed on the lateral side of the upper arm near the elbow.

The orientation of the IMU with respect to the forearm was found as the subject performed a pronation–supination movement, while the palm of the hand faced downwards at the start and end of the measurement. The upper arm was to be held vertically. It is assumed that the angular velocity during pronation is in the direction of the y-axis. By holding the palm of the hand downwards, it is assumed that the z-axis of the forearm coordinate system points in the vertical direction at the beginning and end of each trial. This vertical direction was measured using the 3D accelerometer in the IMU.

The orientation of the IMU coordinate frame (S) to the segment coordinate frame (F) is expressed using a rotation matrix containing the 3 unit vectors of the forearm, expressed in the IMU coordinate system:

$$^{\mathrm{SF}}\mathbf{R} = \left[{}^{\mathrm{S}}\mathbf{x}^{\mathrm{F}} \;\; {}^{\mathrm{S}}\mathbf{y}^{\mathrm{F}} \;\; {}^{\mathrm{S}}\mathbf{z}^{\mathrm{F}}\right] \tag{1}$$

In this text, the coordinate system in which the vector is expressed is indicated by the left superscript, the segment under consideration by the right superscript. The letter U is used to indicate the upper arm segment

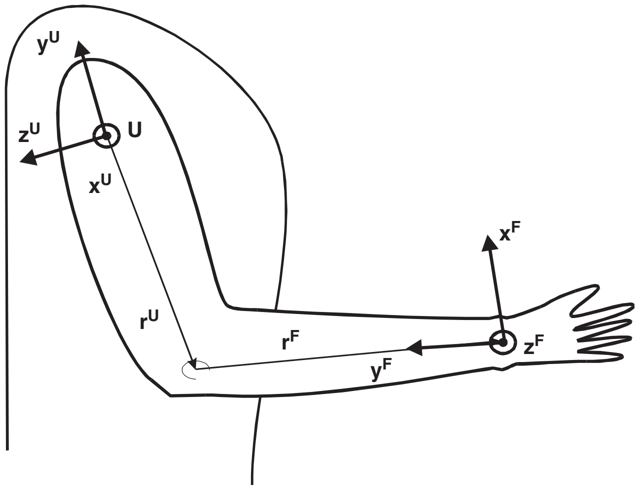


Fig. 1. The definitions of the segment reference frame. The unit y-axis are defined along the segment, upwards. In anatomical position, the z-axis point in dorsal direction and the x-axis laterally.

and associated coordinate system, F is used for the forearm.

The direction of the forearm y-axis can be determined using the direction of the angular velocity during pronation ($\boldsymbol{\omega}_{\mathrm{Pron}}$), or the opposite direction of the angular velocity during supination ($\boldsymbol{\omega}_{\mathrm{Sup}}$):

$$ {}^{S}\mathbf{y}^{F} = \frac{\boldsymbol{\omega}_{\mathrm{Pron}}}{|\boldsymbol{\omega}_{\mathrm{Pron}}|} = -\frac{\boldsymbol{\omega}_{\mathrm{Sup}}}{|\boldsymbol{\omega}_{\mathrm{Sup}}|} \tag{2} $$

Likewise the z-axis can be found by measuring the direction of gravity at the start and end of the trial (Eq. (3)). At the start and end of the trial, the z-axis of the forearm has a direction opposite to the gravity vector. The accelerometer part of the IMU can be used to measure this vector:

$$ {}^{S}\mathbf{z}^{F-} = \frac{-\mathbf{g}_{\mathrm{Start}}}{|\mathbf{g}_{\mathrm{Start}}|} \tag{3} $$

The minus sign in the right subscript is used to denote that it is a first guess, as the z-axis is later recomputed. The x-axis can be found making an orthogonal coordinate system. Because the y and z axes of the segment are defined by measurements, they may not be exactly orthogonal due to measurement errors. The direction of the z-axis is particularly difficult to measure because it is hard to keep the forearm horizontal. In order to compensate for this, the z-axis is recomputed using the y and x axes to make the system orthogonal:

$$ {}^{SF}\mathbf{R} = \left[{}^{S}\mathbf{y}^{F} \times {}^{S}\mathbf{z}^{F-} \quad {}^{S}\mathbf{y}^{F} \quad ({}^{S}\mathbf{y}^{F} \times {}^{S}\mathbf{z}^{F-}) \times {}^{S}\mathbf{y}^{F-} \right] \tag{4} $$

The orientation of IMU with respect to the upper arm was found using the following movements:

1. Place the elbow on a tabletop and perform a endorotation/exorotation movement. Assume the rotation axis is the y-axis.
2. Start in anatomical position. Flex the elbow 90°. Then abduct the upper arm while keeping the elbow fixed. Hold the arm still at the start and end of the movement. The direction of rotation defines the z-axis.

The procedure used to compute the orientation of the upperarm with respect to the IMU is the same as for the forearm, except for the determination of the z-axis of the segment. The direction of the z-axis can be found using the gravity at the start and end of the abduction movement (Eq. (5)). The gravity vector was measured using the 3D accelerometer.

$$ {}^{S}\mathbf{z}^{U-} = \frac{{}^{S}\mathbf{g}_{\mathrm{Start}} \times {}^{S}\mathbf{g}_{\mathrm{End}}}{|{}^{S}\mathbf{g}_{\mathrm{Start}} \times {}^{S}\mathbf{g}_{\mathrm{End}}|} \tag{5} $$

2.2. *Anatomical constraints in elbow*

The elbow of a healthy subject admits flexion/extension and pronation/supination. Abduction/adduction of the elbow is restricted to small angles. It is assumed that the coordinate systems that are identified by the segment calibration can be used to describe this constraint axis. Here we will assume that the y-axis of the forearm will always be in the zy-plane of the upper arm. The adduction angle γ is here defined as the angle between the x-axis of the upperarm and the y-axis of the forearm $-90°$. In radians this can be approximated using the dot product

$$ \gamma \simeq \mathbf{x}^{U} \bullet \mathbf{y}^{F} \tag{6} $$

An estimate of $\mathbf{x}^{U}$ can be obtained by taking the first column of the upper arm rotation matrix. Likewise $\mathbf{y}^{F}$ is the second column of the forearm rotation matrix.

A least-squares filter was designed to use the constraint that the adduction angle is zero to improve the orientation estimate generated using only gyroscopes and accelerometers. Each time step, the orientation of the upperarm and the forearm is estimated using gyroscopes, accelerometers and the previous orientations according to Luinge and Veltink (2005), yielding 2 orientation estimates ${}^{GU}\hat{\mathbf{R}}_t^{-}$ and ${}^{GF}\hat{\mathbf{R}}_t^{-}$ with their variances given by error covariance matrices $\mathbf{Q}_{\theta,t}^{U}$ and $\mathbf{Q}_{\theta,t}^{F}$, respectively. The minus sign is used to indicate an *a priori* estimate, before being corrected. The hat symbol is used to indicate an estimate as opposed to the real value. The least-squares filter estimates the orientation errors ${}^{G}\boldsymbol{\theta}^{F}$ and ${}^{G}\boldsymbol{\theta}^{U}$ in a way that sets the adduction angle to zero. For this purpose a function relating the orientation errors to an adduction angle is required. Finally, the estimated orientation error is used to correct the orientations ${}^{GU}\hat{\mathbf{R}}_t^{-}$ and ${}^{GF}\hat{\mathbf{R}}_t^{-}$ to obtain ${}^{GU}\hat{\mathbf{R}}_t^{+}$ and ${}^{GF}\hat{\mathbf{R}}_t^{+}$, the input of the next step.

In order for the least-squares filter to correct the orientation in a way that sets the adduction angle to zero, a function will be derived that relates the orientation error to the adduction angle. An orientation is described by a rotation matrix. The orientation error is expressed using $\boldsymbol{\theta}$, which has the direction and smallest magnitude that the real orientation of a segment has to rotate in order to coincide with the estimated orientation. For small angles an error of the unit x-axis of the upper arm can be described using the cross product (Bortz, 1971):

$$\mathbf{x} = \hat{\mathbf{x}} - \hat{\mathbf{x}} \times \boldsymbol{\theta} \tag{7}$$

Using the relation

$$\mathbf{x} \times \boldsymbol{\theta} \bullet \mathbf{y} = \mathbf{y} \bullet \mathbf{x} \times \boldsymbol{\theta} = \mathbf{y} \times \mathbf{x} \bullet \boldsymbol{\theta} = (\mathbf{y} \times \mathbf{x})^{\mathrm{T}} \cdot \boldsymbol{\theta}$$

and neglecting products of errors, the relation was found describing the estimated adduction $\hat{\gamma}$ as a function of the real adduction γ and orientation errors:

$$\begin{aligned}\hat{\gamma} &= {}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \bullet {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}} \\ &= \left({}^{\mathrm{G}}\mathbf{x}^{\mathrm{U}} + {}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \times {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{U}}\right) \bullet \left({}^{\mathrm{G}}\mathbf{y}^{\mathrm{F}} + {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}} \times {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{F}}\right) \\ &= \gamma + {}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \bullet {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}} \times {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{F}} + {}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \times {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{U}} \bullet {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}} \\ &= \gamma + \left({}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \times {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}}\right)^{\mathrm{T}} \cdot {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{F}} + \left({}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}} \times {}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}}\right)^{\mathrm{T}} \cdot {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{U}} \end{aligned} \tag{8}$$

A small dot is used to describe a matrix multiplication and a larger dot to indicate the dot product. To obtain the orientation errors using a linear least-squares technique, Eq. (8) was written as a matrix multiplication and the real γ was set to zero:

$$\begin{aligned}\hat{\gamma} &= \left[\left({}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \times {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}}\right)^{\mathrm{T}} \;\; \left({}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}} \times {}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}}\right)^{\mathrm{T}}\right] \begin{Bmatrix} {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{F}} \\ {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{U}} \end{Bmatrix} \\ &= \mathbf{H} \cdot \begin{Bmatrix} {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{F}} \\ {}^{\mathrm{G}}\boldsymbol{\theta}^{\mathrm{U}} \end{Bmatrix} \end{aligned} \tag{9}$$

According to Gelb (1999), the optimal estimate of such an equation can be obtained by

$$\begin{Bmatrix} {}^{\mathrm{G}}\hat{\boldsymbol{\theta}}^{\mathrm{F}} \\ {}^{\mathrm{G}}\hat{\boldsymbol{\theta}}^{\mathrm{U}} \end{Bmatrix} = \mathbf{K} \cdot \hat{\gamma} \tag{10}$$

where $\mathbf{K}$ is defined as

$$\mathbf{K} = \mathbf{Q} \cdot \mathbf{H}^{\mathrm{T}} \cdot [\mathbf{H} \cdot \mathbf{Q} \cdot \mathbf{H}^{\mathrm{T}} + R]^{-1} \tag{11}$$

$\mathbf{H}$ and $\hat{\gamma}$ can be entirely calculated using ${}^{\mathrm{G}}\hat{\mathbf{x}}^{\mathrm{U}}$ and ${}^{\mathrm{G}}\hat{\mathbf{y}}^{\mathrm{F}}$, obtained from the a priory orientation estimates. R is the variance of the adduction angle, a measure of the error that is made by the assumption that the adduction angle is zero. $\mathbf{Q}$ is the covariance matrix describing the covariances of the a priori estimated orientation errors:

$$\mathbf{Q} = \begin{bmatrix} \mathbf{Q}_{\theta,t}^{\mathrm{U}} & \mathbf{0} \\ \mathbf{0} & \mathbf{Q}_{\theta,t}^{\mathrm{F}} \end{bmatrix} \tag{12}$$

The estimated orientation errors were expressed as a rotation matrices (Bortz, 1971) and used to correct the orientation.

3. Experimental methods

The method was tested on one subject by comparing elbow orientations obtained using the IMUs with the orientations as determined using a laboratory bound optical motion capturing system (Vicon). The following procedure was used: markers and IMU's (Xsens technologies, Enschede, 3° rms orientation error) were attached to the upperarm and the forearm (Fig. 2). The forearm IMU was placed on the dorsal side of the forearm near the wrist and the upperarm IMU was placed on the lateral side of the forearm near the elbow. Sensor to segment calibrations were conducted. Initial pose was with the arms along the body and the thumbs forward. Every movement required for the calibration was conducted 5 times and averaged. The subject signed an informed consent prior to measurement.

The subject performed 2 tasks: mimicking eating routines and mimicking morning routines. The eating task consisted of the following activities: pouring a glass (10 s), eating soup (20 s), eating spaghetti (20 s), eating meat (30 s), drinking (10 s). The morning routines task consisted of: splashing water on face and drying it using a towel (10 s), applying deodorant (10 s), buttoning a blouse (10 s), combing hair (20 s), brushing teeth (30 s).

The orientation of the upperarm with respect to the forearm was determined using the described method as well as using the Vicon reference system. The error was defined as the magnitude of the angle the estimated forearm orientation had to be rotated in order to coincide with the forearm orientation obtained using Vicon. The resulting error was compared to the error that was obtained by using the orientation, estimated without using the relation between upperarm and forearm. The orientation of the Vicon marker frame with respect to the segment was obtained using the same procedure as for the IMU's.

Fig. 2. Attachment of Vicon markers and inertial sensing units to the forearm and upperarm. Sensors are adhered to the segment using double sided adhesive tape and secured using Leukoplast.

4. Results

The accuracy of the reference measurements depends on the accuracy of the position measurement of the markers. The accuracy of the position measurement was estimated by considering the distance between 2 markers. The standard deviation of the fluctuation in measured distance was 1 mm. This corresponds to a standard deviation in marker-frame orientation of less than 1°.

The angular velocity in the sensor and segment coordinate frames during endo/exorotation around the upper arm y-axis is shown in Fig. 3. The IMU was placed on the upper arm with the x-gyroscope along the forearm. After determining the orientation of the IMU with respect to the segment, the angular velocity of the segment could be expressed in the segment coordinate frame, causing the rotation around the x-axis to be transformed to an angular velocity around the y-axis.

An example of the performance of the method is shown in Fig. 4. The error was defined as the smallest angle about which the estimated orientation of the forearm with respect to the upperarm has to be rotated in order to coincide with the forearm/upperarm orientation obtained by the reference system. The error obtained using the method described in this study was compared to the error obtained by the method without using the elbow constraint. For the graphs shown in Fig. 5, the standard deviation of γ was set to 10°. It can be seen that although both errors are still considerable, the orientation estimate obtained using the elbow constraint is much smaller than the orientation estimate that does not use this assumption.

The assumption that the adduction angle is zero was tested using the video camera reference system. Fig. 5 shows the adduction angle during a morning routine task. The rms value of the angle was 8°. The least-squares algorithm for estimation of orientation errors requires the specification of R (see Eq. (11)), being the variance of the adduction angle. If R is set close to zero, it is to be expected that the filter will adjust the orientation such that the adduction angle is close to zero. For the 2 measured tasks, the rms value of the calculated adduction angle is plotted for several values of R (Fig. 6a). For stricter values of the elbow constraint, the rms of the estimated adduction angle diminishes. However, this does not necessarily result in a better orientation estimate, as compared to the Vicon reference measurement (Fig. 6b).

5. Discussion

The results clearly indicate that the method to impose anatomical restrictions in the elbow does in fact improve the estimate of forearm to upperarm orientation. However, errors are still too large for many practical applications. Two possible explanations for these large errors are described. First of all, it was assumed that the arms can be described by rigid segments with a single well-defined adduction axis. In reality, there will be movement artefact and the momentary axis will change

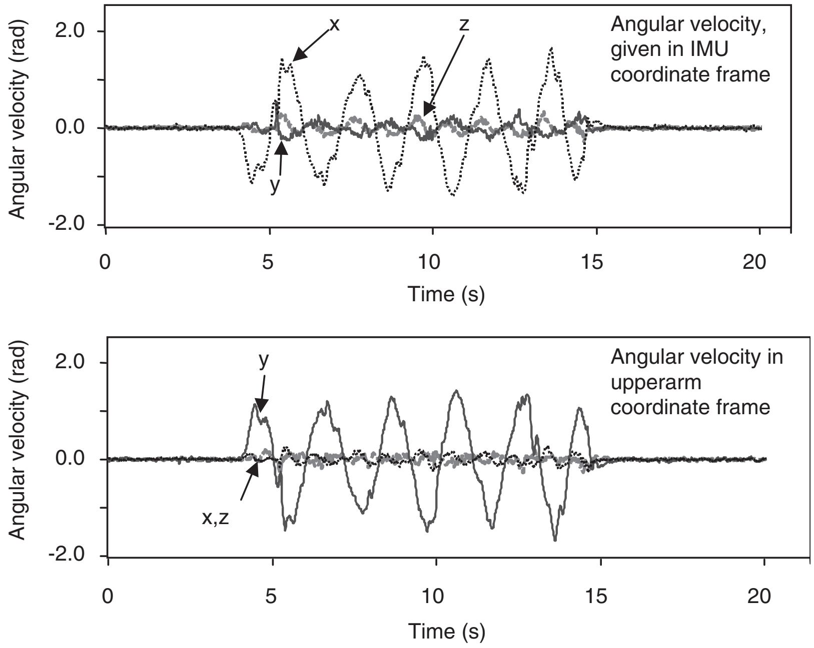


Fig. 3. Angular velocity as measured in the coordinate frame of the IMU during an endorotation/exorotation movement (top graph). This endorotation/exorotation movement was used to find the y-axis of the upper arm. The same angular velocity, now expressed in the upperarm coordinate frame, is given in the bottom graph.

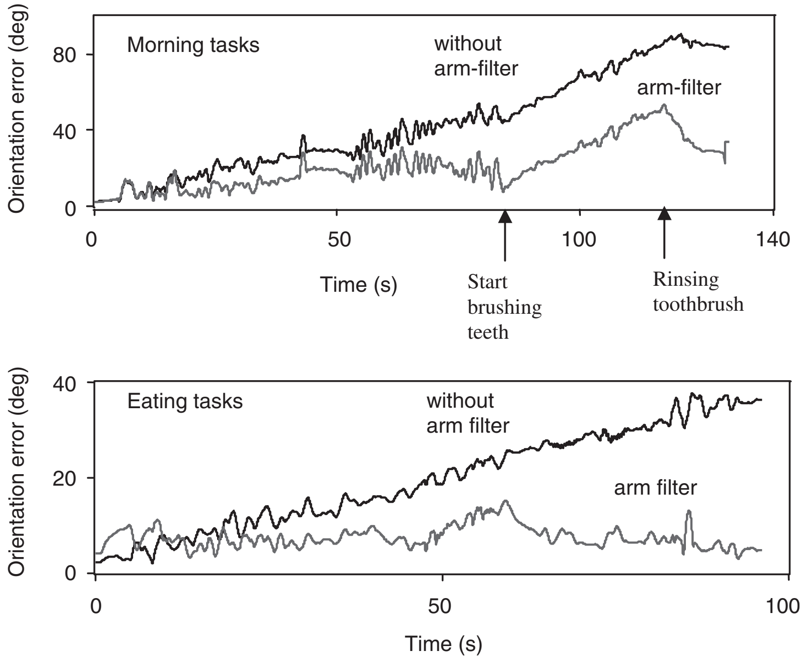


Fig. 4. The error of the orientation of the upperarm with respect to the forearm for a morning routine task and an eating routine task, obtained using the method described in the text (arm filter). The error is compared to the case in which the orientation is not corrected using the constraint on the abduction angle (without arm filter).

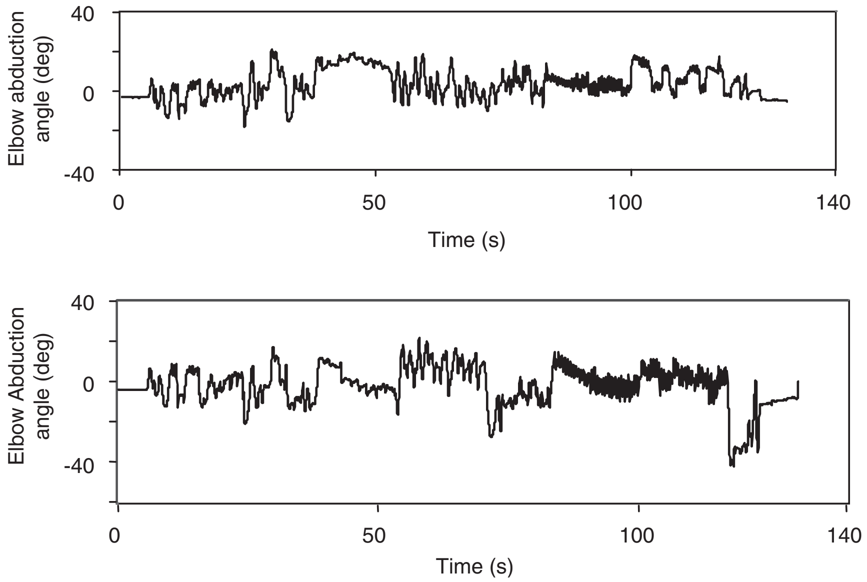
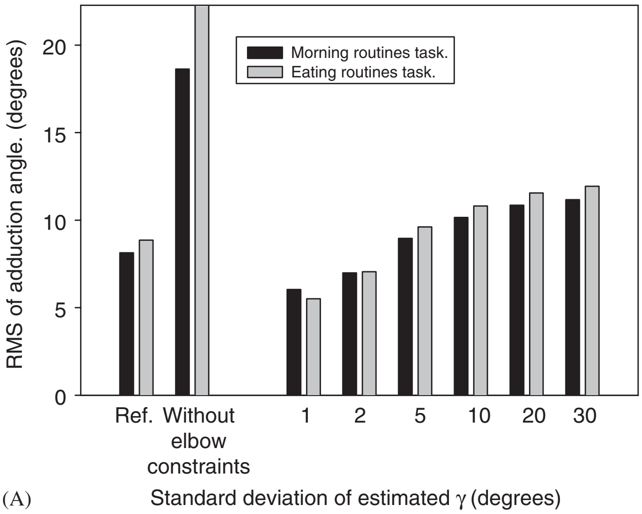


Fig. 5. Adduction angle as measured using the reference system (top) and inertial sensors (bottom) during a morning routine task. The adduction angle in the bottom graph was found using a standard deviation of the elbow abduction angle (square root of R) of 10°.

with the elbow angle. Secondly, the segment calibration may not be accurate. This calibration is not only intended to formally define a coordinate system, but also to approximate a physiological constraint axis. The assumption that the movements can characterize this axis will therefore have limited accuracy.

The described sensor-to-segment calibration is a practical method to estimate the orientation of the IMU with respect to the segment, thereby removing the need for an exact alignment of the IMU on the segment. Most healthy subjects easily perform the movements that are required. The adduction axis is described by the axis perpendicular to the upper arm x-axis and the forearm y-axis. Therefore, these are the 2 axes which are most important to be determined. Errors in estimated segment axes will cause elbow flexion and pronation/



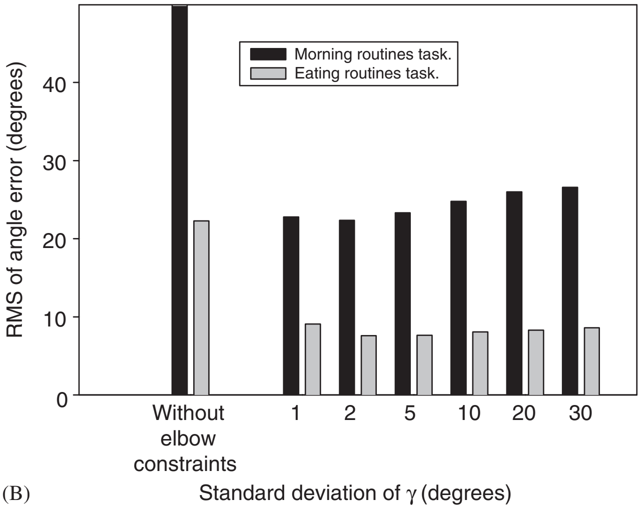


Fig. 6. (A) RMS of estimated adduction angle during a morning routines trial and an eating routines trial. The trials were processed for several values of the model parameter R, the assumed variance of the elbow adduction angle, expressed in the figure as a standard deviation. For comparison, the RMS of the abduction angle obtained with the reference frame is given, as well as the RMS for the situation in which only the gyroscopes and accelerometers are used without assumptions concerning the elbow angle. The estimated adduction angle is smaller for lower values of the standard deviation of γ, (B) RMS errors of elbow orientation without and with the application of elbow constraints at several values of the standard deviation of γ (square root of model parameter R). The errors were determined with respect to the Vicon reference orientations.

supination to be conceived as an apparent elbow adduction. Whether or not the errors in the proposed sensor to segment calibration method were structural, as well as to what extent the method will be different for different subjects were not tested.

An erroneous arm model could also result in orientation estimation errors. In reality, neither the segments are rigid nor the adduction axes can be described by 1 single axis which does not move with respect to the segment (Prokopenko et al., 2001). Especially the IMU on the upperarm is difficult to attach rigidly.

The authors think that improvements of the method could be made by improving the sensor to segment calibration. Given a better segment to sensor orientation and identification of the adduction axis, the anatomical constraint can be relied on more heavily. This will result in a heading adjustment that is accurate even when the arm is nearly horizontal. Improvements could be made by performing more and different calibration movements or having the movements conducted along a board or a wall.

The last possible cause for the errors is the limited observability of the upperarm to forearm orientation using the elbow adduction angle. Inertial sensors as used according to the method using gyroscopes and accelerometers only yield an accurate inclination estimate, but suffer from a slow integration drift around the vertical. It is this drift in heading angle that has to be estimated using the elbow constraint. If the adduction axis, which is the axis perpendicular to the upperarm x-axis and forearm y-axis, is nearly vertical, a heading error will cause a change in estimated adduction, which can be corrected. This can be seen in Fig. 5, where the orientation error increases during the brushing of the teeth. During this movement, the forearm was almost horizontal, resulting in orientation drift. As soon as the arm is lowered, the heading angle was adjusted.

The current study demonstrates that the application of the adduction constraint on the elbow angle can result in considerable improvement in the estimated relative orientation of upperarm and forearm. General statements regarding accuracy of the sensor to segment calibration and of the orientation estimate cannot be made, because the method was tested for only two trials on the same subject. However, the results give an indication of the accuracy that can be expected. It should be noted that orientation errors in the order of 20° are large for some applications such as internal load estimation using inverse dynamics. For other applications such as the assessment of activities of daily living, the accuracy may be sufficient and therefore the described segment calibration procedure and orientation measurement method can attribute to a relevant characterization of arm tasks.

Acknowledgement

The support of the Dutch Technology Foundation STW (contract TEL. 4164) is gratefully acknowledged.

References

Bachman, E.R., 2000. Inertial and magnetic tracking of limb segment orientation for inserting humans in synthetic environments. Ph.D. Thesis, Naval Postgraduate School, Monterey.

Baten, C.T.M., Watermulder, M., Magermans, D., Luinge, H., Koopman, B., 2002. Ambulatory spinal curvature estimation applying 3D motion analysis through inertial sensing for use in net spinal moment estimation. In: Seventh Proceedings of the International Symposium on the 3D Analysis of Human Movement, Newcastle upon Tyne.

Beer, R.F., Dewald, J.P.A., Rymer, W.Z., 2000. Deficits in the coordination of multijoint arm movements in patients with hemiparesis: evidence for disturbed control of limb dynamics. Experimental Brain Research 131, 305–319.

Bernmark, E., Wiktorin, C., 2002. A triaxial accelerometer for measuring arm movements. Applied Ergonomics 33 (6), 541–547.

Bortz, J.E., 1971. A new mathematical formulation for strapdown inertial navigation. IEEE Transactions of Aerospace and Electronic Systems 7, 61–66.

Dunnewold, R.J.W., Jacobi, C.E., van Hilten, J.J., 1997. Quantitative assessment of bradykinesia in patients with Parkinson's disease. Journal of Neuroscience Methods 74, 107–112.

Foxlin, E., Harrington, M., Altshuler, Y., 1998. Miniature 6-DOF inertial system for tracking HMDs. In: Helmet and Head-Mounted Displays III, Aerosense 98, Orlando, FL.

Gelb, A., 1999. Applied Optimal Estimation. MIT Press, Cambridge, MA.

Goldvasser, D., McGibbon, C., Krebs, D., 2001. High curvature and jerk analyses of arm ataxia. Biological Cybernetics 14, 745–752.

Hansson, G.Å., Balogh, I., Ohlsson, K., Rylander, L., Skerving, S., 1996. Goiniometer measurement and computer analysis of wrist angles and movement applied to occupational repetitive work. Journal of Electromyography and Kinesiology 6, 23–35.

Hoff, J.I., Hilten, B.J.V., 1999. A review of the assessment of dyskinesias. Movement Disorders 15, 737–743.

Kemp, B., Janssen, A.J.M.W., van der Kamp, B., 1998. Body position can be monitored in 3D using miniature accelerometers and earth-magnetic field sensors. Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control 109, 484–488.

Luinge, H.J., Veltink, P.H., 2004. Inclination measurement of human movement using a 3D accelerometer with autocalibration. Trans. on neural systems and rehabilitation engineering. IEEE Transactions on Neural Systems and Rehabilitation Engineering 12, 112–121.

Luinge, H.J., Veltink, P.H., 2005. Measuring orientation of human body segments using miniature gyroscopes and accelerometers. Medical and Biological Engineering and Computing 43, 273–282.

Prokopenko, R.A., Frolov, A.A., Biryukova, E.V., Roby-Brami, A., 2001. Assessment of the accuracy of a human arm model with seven degrees of freedom. Journal of Biomechanics 34, 177–185.

Roetenberg, D., Luinge, H.J., Baten, C.T.M., Veltink, P.H., 2005. Compensation of magnetic disturbances improves inertial and magnetic sensing of human body segment orientation. IEEE Transactions on Neural Systems and Rehabilitation Engineering 13 (3), 395–405.

Topka, H., Konczak, J., Dichgans, J., 1998. Coordination of multi-joint arm movements in cerebrellar ataxia: analysis of hand and angular kinematics. Experimental Brain Research 119, 483–492.

Uswatte, G., Miltner, W.H., Foo, B., Varna, M., Moran, S., Taub, E., 2000. Objective measurement of functional upper-extremity movement using accelerometer recordings transformed with a threshold filter. Stroke 31, 662–667.

van der Helm, F.C., Pronk, G.M., 1995. Three-dimensional recording and description of motions of the shoulder mechanism. Journal of Biomechanical Engineering 117, 27–40.

Willemsen, A.T., van Alste, J.A., Boom, H.B., 1990. Real-time gait assessment utilizing a new way of accelerometry. Journal of Biomechanics 23, 859–863.

Zhu, R., Zhou, Z., 2004. A real-time articulated human motion tracking using tri-axis inertial/magnetic sensors package. IEEE Transactions of Neural Systems Rehabilitation Engineering 12 (2), 295–302.