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Original Article

Compensation for the Effect of Soft Tissue Artefact on Humeral Axial Rotation Angle

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Soft tissue artefact (STA) is caused by the relative displacement of markers or sensors mounted on the skin surface with respect to the underlying bones, and is a major source of error in the kinematic measurement of human movement. In particular, the humeral axial rotation (HAR) is affected by STA. The aim of this study was to propose a method for compensating for STA and to validate its effectiveness. In the proposed method, the HAR angle was calculated by a second-order regression using three independent variables converted from the Cardan angles of the shoulder joint. The calculated HAR angle (HAR-r) was compared with the angle calculated from the direction of the longitudinal axis of the forearm (HAR-f). Highly linear correlations were found between HAR-r and HAR-f when the elbow joint was flexed at 90°. The elbow flexion/extension motion had little influence on the HAR-r, whereas HAR-f became unstable when the elbow joint approached its full extension. Because HAR-r effectively compensates for the STA and is independent of the elbow flexion/extension, the regression method is suitable for the movement analysis of the upper limbs.

Key words: Soft tissue artefact, Movement analysis, Humeral axial rotation angle, Carrying angle

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Introduction

A quantitative description of the upper limbs kinematics is particularly useful and necessary for evaluating the activities of daily living[1,2]. Basically, the rotation angles of the relating joints are estimated from the kinematic data. Errors in this estimation are caused by various factors. Because the accuracy in the measurement equipment is remarkably improved in recent years, soft tissue artifact (STA) has become the major source of errors. STA is caused by the relative displacement of markers or sensors mounted on the skin surface with respect to the underlying bones[3-5].

In particular, the humeral axial rotation (HAR) is most affected by STA[1,3,4,6]. Estimation error in HAR causes deviation of the angles in the shoulder and elbow joints, and biases the range of motion (ROM) of these joints. As a result, functional and clinical evaluation based on the joint motion becomes unreliable. Therefore, in order to obtain an accurate estimate of HAR, it is necessary to apply compensation to the joint angles produced by the markers or sensors attached to the human body.

Leardini *et al.*[5] reviewed the methods for the assessment and compensation of STA especially for the kinematic measurement of the lower limbs. Concerning the upper limbs, there are two major approaches to the compensation of STA[3,4]. The first method was proposed by Schmidt *et al.*[1]. They used two rotation matrices to calculate the orientation of the upper-arm. The first matrix is based on the direction of the forearm. When the elbow joint is flexed less than 15°, the rotation matrix is switched to the second one derived from the markers on the upper-arm. This method may cause a discontinuity when the elbow is flexed and extended across the switching border.

The second method was described by Roux *et al.*[6], who used the global optimisation method developed by Lu and O'Connor[7]. They compensated for the STA in the upper-arm and the forearm as a whole by fitting the measured positions of the markers to a constrained model of the human body. However, because HAR in this method is affected by the direction of the forearm, it may also become unstable when the elbow joint approaches its full extension. Recently, Cutti *et al.*[4] proposed another method based on a regression technique, but they could apply their method only when the elbow is flexed more than 30°.

Therefore, the purpose of the present study was to develop a method of compensating for the effect of the STA on the HAR angle over the entire range of the elbow joint angle.

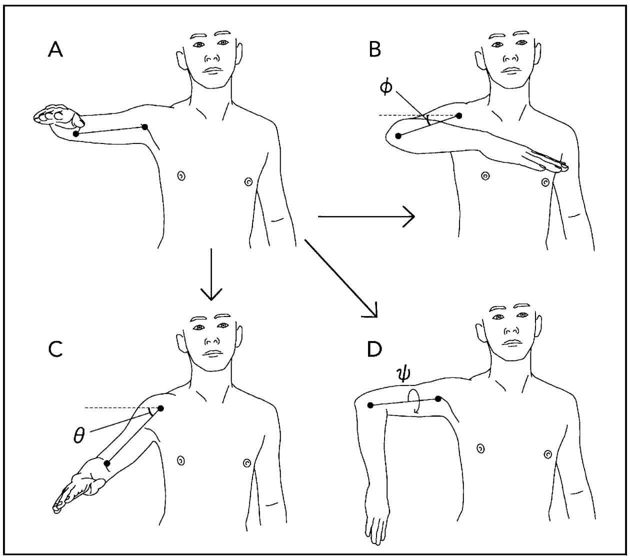

Fig. 1. Description of Cardan angles in glenohumeral joint. A→B: the change in the plane of elevation, A→C: the amount of elevation, A→D: the axial rotation of the upper-arm.

Materials and Methods

Kinematic Model

A rigid-body model consisting of three segments (trunk, upper-arm and forearm) connected by the glenohumeral and elbow joints was used. The glenohumeral joint was modeled as an ideal ball and socket joint with three rotational degrees of freedom (DoF): horizontal flexion/extension, elevation and axial rotation. The elbow joint had two DoFs: flexion/extension and pronation/supination. Joint rotations were described by the Euler/Cardan angles according to the recommendation from the International Society of Biomechanics[8].

For the glenohumeral joint, three Cardan angles φ, θ, ψ defined the plane of elevation, the amount of elevation, and axial rotation, respectively (Fig.1)[9,10]. The angle θ is 90° when the upper-arm is directed downward. To make the amount of elevation 0° at this position, θ was transformed into $\beta=90°-\theta$. Moreover, because the angles φ and ψ became unstable at this position (gimbal lock phenomenon), they were converted to $\alpha=\varphi\cos\theta$, $\gamma=\varphi\sin\theta-\psi$.

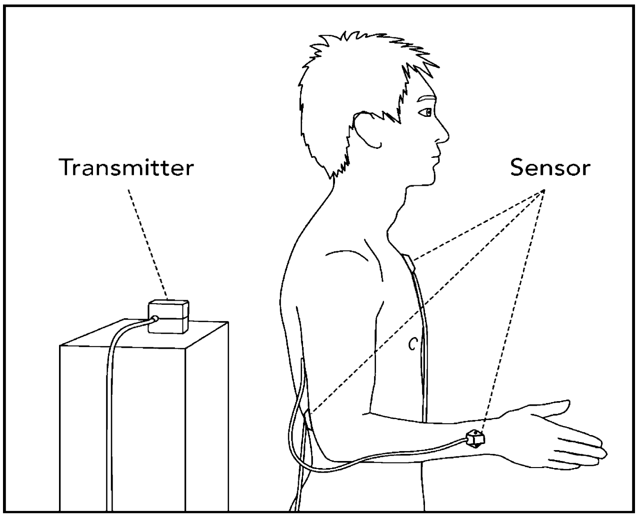

Fig. 2. Configuration of the measurement equipment. The transmitter was placed on a wooden stand behind the subject and three sensors were attached to the sternum, the dorsal side of the distal upper-arm and the most distal part of the forearm.

Recording of Rotation Angles

We used a three-dimensional electromagnetic tracking system consisting of a transmitter, three sensors and a controller unit (Fastrak, Polhemus, U.S.A.)[11]. The three sensors were attached to the sternum, the dorsal side of the distal upper-arm and the most distal part of the forearm with adhesive tape (Fig.2).

The rotation angles of the sensors with respect to the transmitter were measured with a sampling rate of 30 Hz and were transferred to a personal computer. The acquisition of data and the calculation of the angles were executed by custom-made software. The rotation angle γ calculated from the output of the sensor mounted on the upper-arm is denoted by HAR-s in this paper.

HAR Angle Calculated from the Longitudinal Axis of the Forearm

When the elbow joint is flexed at 90°, the HAR can be reliably estimated from the longitudinal axis of the forearm (the forearm method)[4]. The HAR calculated using the forearm method is denoted by HAR-f. It is free from the influence of STA, because the forearm direction is decoupled from the axial rotation of the sensor on the upper-arm[1,3,4]. HAR-f, however, has a fundamental drawback in that it becomes completely unreliable when the elbow approaches its full extension.

To calculate HAR-f, we must take the carrying angle (CA) into consideration. CA is defined as the angle between the forearm and the extension of the upper-arm with the arm fully extended (Fig.3)[12,13]. In this study, CA was measured with a goniometer.

Procedure for Compensation

We proposed a method for estimating HAR using a regression (the regression method). The three angles α, β, γ were used as independent variables, and HAR-f at the elbow flexion of 90° as a dependent variable. We used the following second-order regression

$$\zeta=c_0+c_1\alpha+c_2\beta+c_3\gamma+c_4\alpha^2+c_5\beta^2+c_6\gamma^2+c_7\alpha\beta+c_8\alpha\gamma+c_9\beta\gamma.$$

To determine the 10 regression coefficients c_i (i=0-9), we minimized the root mean square (RMS) of difference between ζ and HAR-f measured during a calibration motion that will be described later. After the determination of c_i, HAR was calculated using ζ. The resulting HAR is denoted by HAR-r.

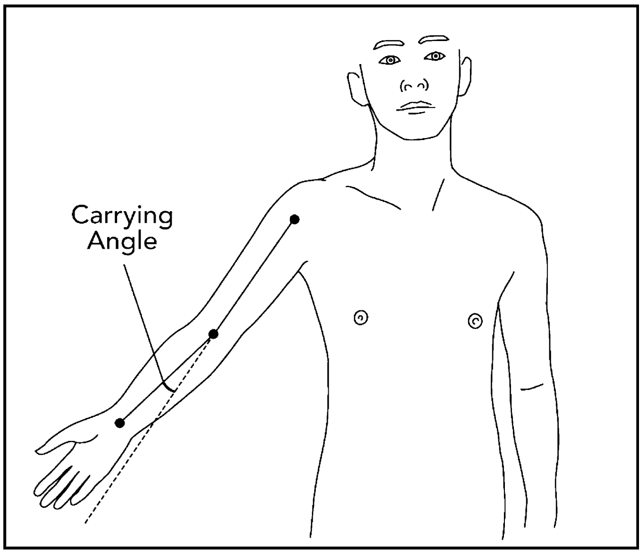

Fig. 3. Definition of the carrying angle.

Subjects and Analyzed Motions

The method was tested with twelve healthy adults (7 males and 5 females), between 21 and 36 years old (average age 27.5±5.6 years). After they received an explanation on the objectives of the study, they gave their informed consent that was approved by the Ethics Committee of Tokyo Medical and Dental University. Only the right side was measured in all subjects.

Prior to the measurements, a reference position was recorded with the trunk upright, the upper-arm hanging aside the trunk, the elbow joint flexed at 90° and the forearm pronation/supination neutral. An orthogonal wooden frame was placed on the side of the upper-arm. The vertical axis of the frame was adjusted with a string hanging downward. The longitudinal axis of the forearm was aligned with a laser spot pointing to a target position marked on a wall.

In order to obtain the regression coefficients, a calibration motion was measured. It consisted of the upper-arm internal/external rotation at 0°, 30°, 60° and 90° elevations of the upper-arm in the planes of horizontal flexion of 0°, 45° and 90° along with keeping the elbow at 90° and the pronation/supination of the forearm neutral.

To clarify the effect of the elbow joint angle on the estimation of the HAR, the elbow was flexed and extended 2-3 times in the range of 0°-maximum flexion with the HAR at 0°, the upper-arm elevation at 30° in the frontal plane and the pronation/supination of the forearm neutral. During this motion, the subjects kept the HAR as constant as possible without moving the upper-arm.

Results

Accuracy of HAR Estimation

We compared the HAR-s and HAR-f as well as HAR-r and HAR-f measured during the calibration motion. Fig. 4 shows a typical example. The range of the HAR-s was smaller than that of the HAR-f due to the influence of the STA (Fig. 4a). Moreover, the zero point of the HAR-s disagreed with that of HAR-f. In contrast to the HAR-s, the HAR-r was highly correlated with HAR-f (Fig. 4b). Because HAR-f should represent the correct HAR with the elbow joint flexed at 90°, HAR-r also reproduced the correct HAR.

Similar tendencies were observed in all the subjects.

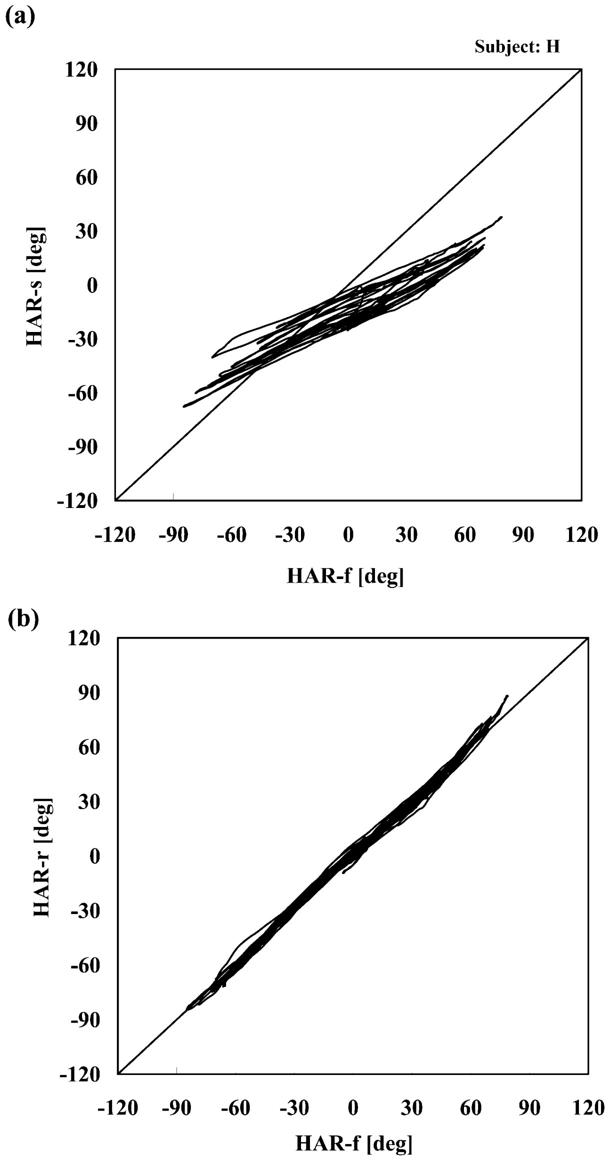

Fig. 4. A typical result of correlations of (a) HAR-s (humeral axial rotation directly calculated from the output of a Fastrak sensor) and (b) HAR-r (by the regression method) with HAR-f (by the forearm method) measured during the calibration motion. The correlation between HAR-s and HAR-f deviated from the 45° diagonal line (a), while HAR-r was highly correlated with HAR-f (b).

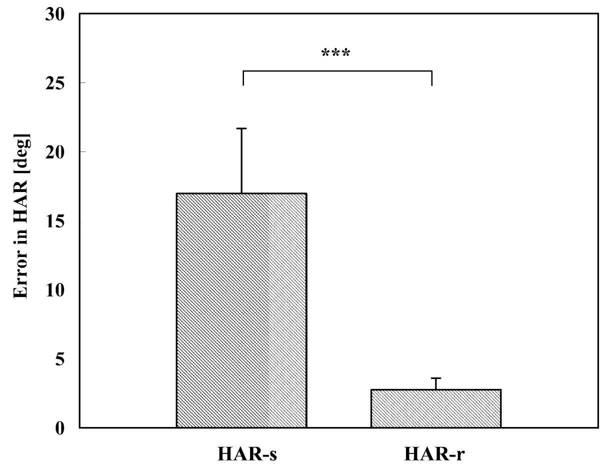

Fig. 5. Mean and standard deviation of RMS for HAR-r and HAR-s calculated for 12 subjects. The RMS for HAR-s was significantly greater than that for HAR-r. ***: $p<10^{-6}$, paired t-test.

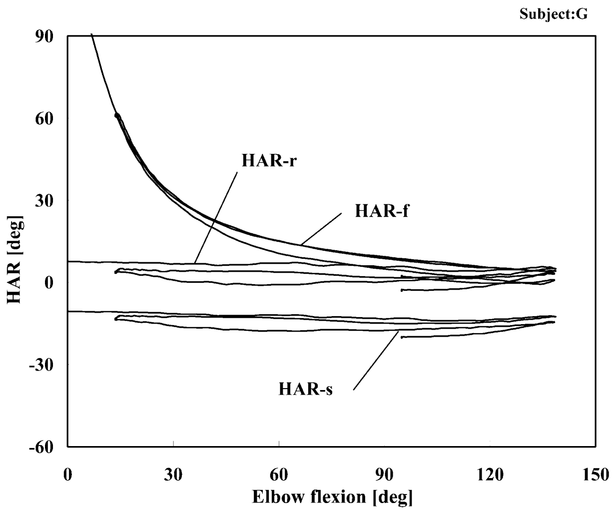

Fig. 6. Influence of elbow flexion/extension on HAR-s, HAR-f and HAR-r while the subject maintained the HAR angle at 0° as accurately as possible. HAR-s deviated from the target HAR of 0°. HAR-f was about 0° at the elbow angle of 90°, but it showed a significant variability depending on the elbow flexion/extension. HAR-r showed the target HAR of 0° and was not affected by the elbow flexion/extension.

The CA used for the calculation of HAR-f varied between 6°-20° (mean 10.83°±3.61°). We evaluated the deviations of HAR-s and HAR-r from HAR-f with the RMS of their differences. Fig. 5 shows the mean and standard deviation (SD) of the RMS for the 12 subjects. The RMS for HAR-s was 16.96°±4.72° and was significantly greater than that for HAR-r (2.76°±0.84°, $p<10^{-6}$, paired t-test).

Effect of Elbow Flexion/Extension

Fig. 6 shows typical changes in the HAR angle when the elbow was flexed and extended. HAR-s was undisturbed by the elbow flexion/extension. However, it deviated from the target angle of 0° due to the influence of STA. HAR-f was about 0° when the elbow was flexed at 90°, but it showed a significant variability depending on the elbow flexion/extension, although the subject kept the HAR constant as accurately as possible.

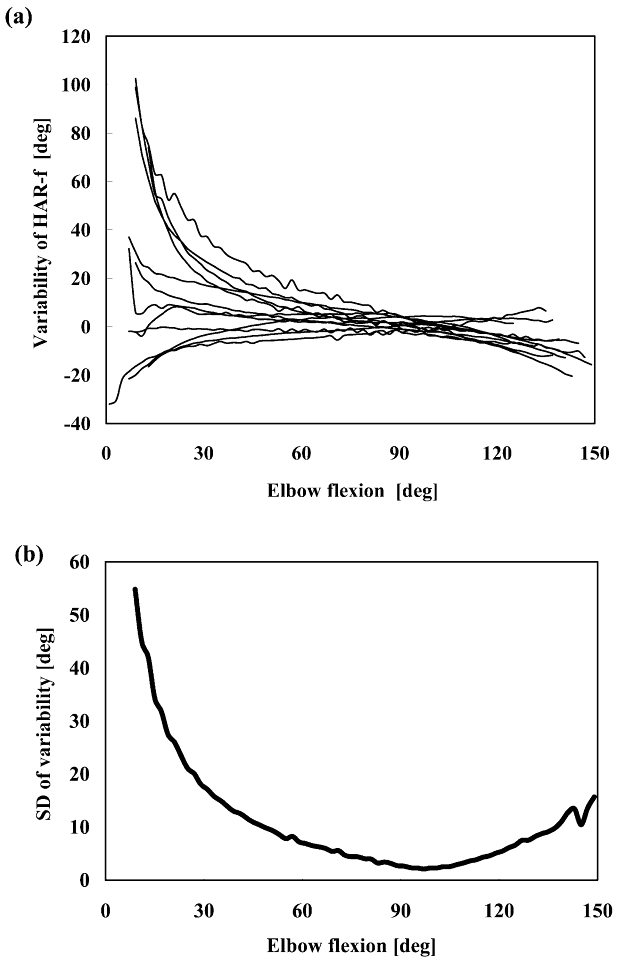

Fig. 7. (a) Variability in HAR-f after subtracting HAR-r depending on the angle of the elbow flexion/extension. Each line represents HAR-f measured from an individual subject.
(b) Standard deviation of the variability of HAR-f shown in (a).

HAR-r showed the correct HAR as HAR-f at the elbow flexion of 90°. Moreover, it was not affected by the elbow flexion/extension. Similar results were obtained in all the subjects.

Because HAR-r is assumed to be independent of the elbow flexion/extension, it should represent the correct HAR angle over the entire range of the elbow flexion. Therefore, we calculated the deviation in HAR-f from HAR-r to evaluate the variability of HAR-f for the 12 subjects. In Fig. 7a, each line represents the deviation of HAR-f measured from each individual subject. The minimal deviation was -31.94° at the elbow angle of 1° and the maximum value was 102.50° at the elbow angle of 9°. The SD of the variability was calculated for all the subjects and is shown in Fig. 7b. The minimal SD was 2.08° at the elbow flexion of 97°. When the elbow joint approached the full extension, HAR-f significantly deviated from 0°. For example, the SD was 54.81° at the elbow angle of 9°.

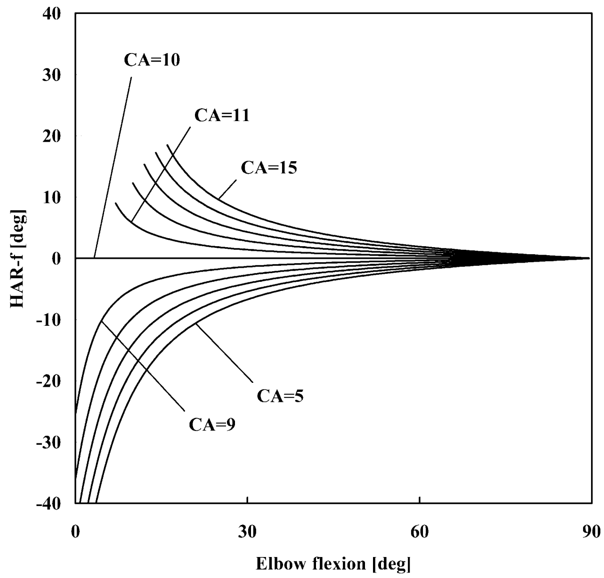

Fig. 8. Theoretical results of the variability of HAR-f depending on the deviation during the setting of the CA. The correct CA was 10°. The HAR-f was aligned with the 0° line when the correct CA was set. If CA was assumed to be greater than the correct value, HAR-f deviated at positive angles. If CA was smaller than the correct value, HAR-f showed negative angles.

Effect of CA

Fig. 8 shows a theoretical influence of the CA on HAR-f. The correct CA was set to 10°. The CA assumed for the calculation of HAR-f was varied every 1° between 5° and 15°. If incorrect CAs were assumed, HAR-f deviated from 0° depending on the elbow flexion/extension. If CA was lower than the correct value, HAR-f showed negative angles. When CA was 1° smaller, HAR-f deviated up to -25.39° at the full elbow extension. If CA was assumed to be greater than the correct value, HAR-f deviated at positive angles. Because the angle between the longitudinal axes of the upper-arm and the forearm cannot be smaller than the correct CA, HAR-f is not defined for the full elbow extension. When CA was 1° greater than the correct value, HAR-f reached 16.33° at the elbow flexion of 5°.

Discussion

HAR-f based on the direction of the forearm longitudinal axis is not affected by STA, because the forearm direction is decoupled from the axial rotation of the sensor on the upper-arm. However, it depends on the degree of the elbow flexion/extension (Figs. 6 and 7) and pronation/supination[3]. In this study, we focused on the effect of the elbow flexion/extension and neglected that of pronation/supination by keeping the axial rotation of the forearm at a neutral position.

Various compensation techniques based on the forearm method had been reported. The methods proposed by Schmidt *et al.*[1] and by Roux *et al.*[6] can be classified as derivatives of the forearm method. Schmidt *et al.* calculated HAR by the forearm method when the elbow joint is flexed more than 15°. When the elbow is extended beyond this angle, the forearm method is abandoned and a second rotation matrix is used in conjunction with the last reliable value of the first rotation matrix in order to calculate the orientation of the upper-arm. This method may reduce the effect of STA, but may cause a discontinuity if the elbow is flexed and extended across the switching border. Roux *et al.* used the global optimization method to minimize the effect of STA. In their method, the orientation of the upper-arm is calculated from the entire set of markers attached to the body. Therefore, the orientation of the forearm is included in the HAR calculation. The contribution of the markers is weighted depending on the susceptibility of the markers to STA. If the contribution of the upper-arm markers to the calculation of HAR is very low, this method may become identical to the forearm method. It may then become unreliable near the full extension of the elbow joint. Cutti *et al.*[3,4] described that the HAR based on the forearm method becomes poorly estimated because the longitudinal axes of the upper-arm and the forearm are almost aligned when the elbow is flexed less than 15°.

The existence of CA makes the estimation based on the forearm method more complex. It should be noted that even after we took the effect of the carrying angle (CA) into consideration, HAR-f was still unstable near the full extension of the elbow joint. As shown in Fig. 7, the deviation of HAR-f from the correct value was in either the positive or negative directions. It is caused by the incorrect setting of the CA for the calculation of HAR-f. However, as shown in Fig. 8, only a 1° of inaccuracy of CA causes a substantial deviation in the HAR. Therefore, even if CA is adjusted to the correct value so that HAR-f is not affected by the elbow flexion/extension, the pronation/supination of the forearm changes the direction of the longitudinal axis of the forearm and alters the CA from the correct value. Consequently, for practical measurements, it is impossible to obtain a CA accurate enough to calculate a reliable HAR-f. Therefore, the forearm method and its derivatives are inappropriate for the compensation of STA.

In our regression expression, we used α, β and γ (=HAR-s) as independent variables. These variables are derived from the Cardan angles of the shoulder joint and consequently are independent of the elbow flexion/extension. Therefore, HAR-r calculated as the dependent variables of the regression is also independent of the elbow motion. Moreover, the regression method is reliable by reproducing HAR-f at the elbow flexion of 90°, where HAR-f represents the correct HAR. Compared to HAR-s, HAR-r reduced the RMS error from 16.96° to 2.76° (Fig. 5). Considering that the second-order regression had reproduced an accurate HAR, it is not necessary to use a regression with third or higher order polynomial expressions.

In Fig. 6, the angle of the HAR-s was negative, while the target HAR was 0° and the upper-arm was elevated at 30°. According to our preliminary trials, the deviation was not found at the upper-arm elevation of 0° and was more pronounced at the higher elevations up to 90°. Because the Fastrak sensor was mounted on the dorsal side of the upper-arm, gravity should cause the sensor to rotate externally around the longitudinal axis of the upper-arm. This external rotation causes the negative angle of HAR-s. It should be noted that this kind of artefact is also compensated by the regression method, as long as the trunk is in the upright position.

Cutti *et al.*[4] proposed a regression method similar to the one in this paper. However, they used the flexion/extension angle of the elbow joint and the pronation/supination angle of the forearm as independent variables in the regression. As a result, they could not apply the method to an elbow flexion range lower than 30°, perhaps because the regression becomes unstable in this range. In the present study, the HAR angle over the entire range of elbow flexion can be corrected and is not affected by the elbow flexion/extension motions. Therefore, the regression method proposed in this paper is a stable and effective means to compensate for the STA.

In addition to γ (=HAR-s), we used the angles α and β (elevation of the upper-arm) as the independent variables during the regression. The original angles φ and ψ are unstable at the gimbal lock position (the upper-arm directed downward). Therefore, if φ and ψ are used in the regression, the regression also becomes unstable. Because we need regression coefficients that can be used over the entire range of the upper-arm motion, we converted φ and ψ into the stable angles of α and γ. With this conversion, the compensation for STA by the regression method becomes possible. Masuda *et al*. proposed to use the angle γ as a new definition of HAR [14], which can replace the conventional definition based on the Cardan angle.

Conclusion

The HAR angle was accurately estimated over the entire motion range of the upper-arm and the elbow joint using a second-order regression and a new definition of HAR. In this method, a calibration motion must be performed for obtaining the regression coefficients. Once the regression coefficients were determined, the compensated HAR angle can be obtained for real-time. Because there are no constraints in the range of the analyzed motion, this method can be applied to a kinematic analysis of the upper limbs during various activities of daily living.

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