Direct Kinematic Modeling of the Upper Limb During Trunk-Assisted Reaching

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The study proposes a rigid-body biomechanical model of the trunk and whole upper limb including scapula and the test of this model with a kinematic method using a six-dimensional (6-D) electromagnetic motion capture (mocap) device. Large unconstrained natural trunk-assisted reaching movements were recorded in 7 healthy subjects. The 3-D positions of anatomical landmarks were measured and then compared to their estimation given by the biomechanical chain fed with joint angles (the direct kinematics). Thus, the prediction errors was attributed to the different joints and to the different simplifications introduced in the model. Large (approx. 4 cm) end-point prediction errors at the level of the hand were reduced (to approx. 2 cm) if translations of the scapula were taken into account. As a whole, the 6-D mocap seems to give accurate results, except for pronosupination. The direct kinematic model could be used as a virtual mannequin for other applications, such as computer animation or clinical and ergonomical evaluations.

Keywords: trunk-assisted reaching, direct kinematics, coordination, 6-D motion capture, model

During the last decade, many studies have developed the use of electromagnetic motion capture (mocap) for the kinematics of the upper limb (Meskers et al., 1998; Fayad et al., 2008; Davardhini et al., 2005). The general methodology relies on rigid bodies modeling. Experimentally, an electromagnetic sensor is fixed on each body segment and representative bony landmarks are located in the reference frame of each corresponding sensor. Then the bony landmarks are used for the definition of the local coordinate system of each segment and for the computation of joint rotations according to the recommendations of the ISB standardization proposal (Van der Helm, 2002; Wu et al., 2005). Validation studies with bone-fixated sensors demonstrated that the method with electromagnetic sensors was reliable when arm elevation remained below 120° (Karduna et al., 2001; Ludewig et al., 2009). This method has good reliability and accuracy (Crosbie et al., 2008; Lovern et al., 2009) and is now widely used to measure the mobility of the shoulder complex during arm elevation (Amasay et al., 2009; Rundquist et al., 2009; Vermeulen et al., 2002). The use of electromagnetic sensors is also validated to measure the kinematics of the upper limb (Biryukova et al. 2000; Prokopenko et al. 2001). However, to our knowledge, such a method has still not been validated for the combined analysis of the trunk, shoulder complex, and upper limb during large functional movements, such as trunk-assisted reaching with the target beyond the anatomical length of the arm. The aims of our study are (1) to formalize a direct kinematic model of five rigid body segments (trunk, scapula, upper arm, forearm, hand) built from electromagnetic mocap of the upper limb and linked by joints that have three angular degrees of freedom; (2) to estimate the errors coming from modeling simplifications, that is, rigid bodies and ideal joints (Figure 1); and (3) to experimentally measure the contribution of scapular movement (rotation and translation) to the displacement of the hand during trunk-assisted reaching movements.

Methods

Instrumentation and Recording of Bony Landmarks

Position and orientation of six sensors are recorded at a sampling frequency of 86 Hz by an electromagnetic tracking device (Motion Star, Ascension Technology Corp.). The accuracy is 0.5 mm RMS and about 0.01° for rotations (Milne et al., 1996; Poulin & Amiot, 2002). Sensors were fixed on the skin with double-sided adhesive tape over each segment respectively at the level of the sternal manubrium for the thorax; on the flat surface of the superior acromion process for the scapula; strapped to the lateral arm just below the insertion of the deltoid for the upper arm; on the posterior surface of the lower arm, roughly 8–10 cm above the wrist level; and on the posterior surface of the hand with the main axis of the sensor along the third metacarpal bone (Figure 2).
Figure 1 — Procedure for data acquisition and modeling (thick arrows). The aim of the study is to compare the trajectories of bony landmarks (i) directly measured and (ii) computed from the estimation of the joint angles (the “direct” model). The different sources of errors are indicated (double arrowheads). The present procedure estimates the errors due to model inadequacies but not other human or technical measurement errors.

Figure 2 — Experimental and recording setup. The task was to put the hand on a starting position (A), to reach and grasp a cylindrical object (B), and to lift it onto a shelf (C). The numbers 1–5 indicate the placement of the sensors on the trunk (5), scapula (1), upper arm (2), lower arm (3), and hand (4).
The sixth sensor, mounted on a pointer, was used to digitize at rest the position of the following anatomical bony landmarks in the reference frame of the corresponding sensor: processus xiphoideus, incisura jugularis, and processus spinosus of C7 and T8 for the trunk; angulus acromialis, trigonum spinae, and angulus inferior for the scapula; medial and lateral epicondyles for the upper arm; and radial and ulnar styloids for the lower arm. The position of the glenohumeral rotation center was estimated from passive circumduction movements of the upper limb (Harryman et al., 1990; Biryukova et al., 2000). The 3-D positions of these anatomical points can then be computed during any movement by means of mocap data and local-to-global transformation. The position of the hand is directly measured by the fifth sensor.

This method provides a description of the upper limb as a set of five polygons limited to four (trunk, forearm) or three (scapula, upper arm, hand) points, giving a total of 17 points described as coordinates of the jth bony landmark of the ith segment.

Biomechanical Hypotheses and Calculation of Joint Angles

The 3-D positions of the bony landmarks were used to define the anatomically based local coordinate system of each segment (Meskers et al., 1998). Each segment is thus defined by (1) a set of anatomical reference points giving its shape, (2) a reference frame, and (3) the origin of the reference frame. The position of the ith segment relative to i−1 is given by a sequence of three Euler angles and the coordinates of the origin of the segment reference frame in the reference frame of the (i−1)th segment. The local coordinate system definition and Euler angle sequence of rotations were consistent with the ISB standardization proposal (Van der Helm, 2002).

Direct Kinematics and Test of the Model

Let the following equation represent the set of the coordinates of the k bony landmarks of the ith segment expressed in its local coordinate system.

\[ X_{i}^{k}_{i} = \left[ X_{i,1}^{k}, X_{i,2}^{k}, \ldots, X_{i,n}^{k} \right] \quad (1) \]

Let \( X_{i,0}^{k} \) be the origin (and the rotation center) of the reference frame of this ith segment expressed in the local coordinate system of the preceding segment in the kinematic chain. And finally, let \( A_i \) and \( M_i \) be, respectively, Euler’s angles and the attitude matrix that determine the orientation of the ith segment in the reference frame of the (i−1)th segment. For each segment, we have the following:

\[ X_{i}^{j} = \left[ X_{i,1}^{j}, X_{i,2}^{j}, \ldots, X_{i,n}^{j} \right] \quad (2) \]

the set of coordinates of k bony landmarks in the global reference frame was computed by applying simple 3-D rotation rules. For instance, for the jth anatomical landmark of the humerus, we have the following:

\[ X_{GCS}^{j} = X_{i}^{j} + M_{j} \cdot X_{i}^{j-1} + M_{j} \cdot M_{j-1} \cdot \left[ X_{i}^{j-1} + M_{j-1} \cdot M_{j-2} \cdot \left[ \ldots + M_{2} \cdot \left[ X_{i}^{1} \right] \ldots + M_{1} \cdot X_{i,0}^{1} \right] \right] \quad (3) \]

In order to test the biomechanical hypotheses founding the model, we computed the distance \( d_{i,j} = |X_{i}^{j} - X_{GCS}^{j}| \) between the measured posture of the subject \( X_{GCS}^{j} \) and the estimate of this posture \( X_{i}^{j} \) given by the direct kinematic model fed with computed joint angles. We focused on several points: the barycenter of the scapula, the center of the elbow and wrist joints (midpoint between the medial and lateral epicondyles and the radial and ulnar styloids, respectively) and the hand. In order to estimate the contribution of the scapular motion, we computed these errors with and without taking into account the translations of the scapula during the movement. To that purpose, the direct upper-limb kinematic chain was built from (1) a static scapula center, computed as the barycenter of the three landmarks palpated at rest and then fixed in the thorax reference frame (fixed vector from the barycenter of the trunk to the barycenter of the scapula) and (2) a scapula center moving as recorded experimentally.

Experimental Procedure

Seven right-handed healthy subjects (5 women and 2 men, 24–46 years old) participated in the study. All subjects were professional colleagues or students who gave a written consent for the experiment, in agreement with French regulations on ethical issues. They were comfortably seated on a chair in front of the table (abdomen at 10 cm from its edge), keeping the right hand on the table. The task was to reach and grasp a cylindrical object (8.4 cm in height, 7 cm in diameter with two different weights: 300 g and 1000 g) placed on the table in front of the subject at a distance equal to the length of his or her maximally extended left arm and to lift it onto a shelf (18 cm in height) located 20 cm in front. The instruction was to “act naturally,” without emphasis on precision or velocity. The onset of the movement is defined when the velocity of the hand exceeds 0.05 m·s⁻¹, time of grasping when it drops below 0.08 m·s⁻¹, and time of deposit when object velocity drops below 0.08 m·s⁻¹. Sixty movements were recorded for each subject.

Results

Kinematics of the Hand

The mean duration of the movement until the release of the object was 1.41 ± 0.10 s (0.67 ± 0.28 s for reaching and 0.75 ± 0.28 s for transport). The displacements of the hand were quite wide: 48.67 ± 1.77 cm during reaching and 0.75 ± 0.28 s for transport). The displacements of the object was 1.41 ± 0.10 s (0.67 ± 0.28 s for reaching and 0.75 ± 0.28 s for transport).
The Upper Limb During Trunk-Assisted Reaching

Figure 3). For the elbow, the first and third elbow Euler angles correspond respectively to extension and supination; however, there was a significant amount of rotation in the second Euler angle (abduction-adduction), which should be anatomically fixed. Similarly, there was a small (less than 5°) but significant rotation in the third wrist angle, along the main axis of the hand during reaching. These findings prompted us to compare (paired t test) the values of the flexion-extension angles to elbow 3-D geometrical angles obtained from vector computation. This geometrical angle is a 3-D angle between two vectors. For the elbow, the vectors represent respectively the upper arm (between the centers of the glenohumeral and elbow joints) and the lower arm (between the centers of the elbow and wrist joints). The cosine of this angle is the dot product of the vectors divided by the product of their norms. There were no significant differences before movement or at the time of grasping, and a slight but significant difference at the time of deposit (3.4° for elbow and 5.6° for wrist, p < 0.05).

Comparison of the Measured and Predicted Values

The distances between the measured and predicted points after reconstruction vary during the movement (Figure 4). The raw errors for the model with a static scapula center (M1 model; white bars on Figure 4) were small just before the movement, even at the level of the hand, where all the potential errors of the direct kinematic reconstruction accumulate (0.26 ± 0.03 cm). Errors increased significantly during the movement: at the time of grasping, they were ~2.5 cm at the level of the scapula and elbow but larger than 4.6 cm at the level of the wrist and hand. At the time of deposit, the errors were ~4 cm at the level of the scapula and elbow and reached over 6 cm at the level of the wrist and hand. The errors were smaller when the model included the translation of the scapula (M2 model; black bars on Figure 4). The remaining errors were less than 0.5 cm at the elbow and ~2.5 cm at the level of the wrist and hand.

The mean excursion of the center of the scapula in the reference frame of the thorax was 2.2 ± 0.61 cm until the time of grasping and 3.86 ± 0.95 cm until the time of deposit with a significant forward and upward translation (p < 0.05 at one-sample Student’s t test). However, the distance between the centers of the trunk and of the scapula varied less than 0.5 cm (ns).

Discussion

Protocols based on magnetic mocap systems and using anatomical body landmarks to compute local reference frames in accordance with the international ISB proposal (Wu et al., 2005) are now largely used to analyze shoulder function (Fayad et al., 2006; Lin et al., 2006; Vermeulen et al., 2002). Here we propose a more general model involving the trunk and the whole upper limb, and we challenged this model with functional gestures involving large hand movements and most of the upper limb degrees of freedom.

Figure 3 — Amount of joint rotation during reach (light gray) and lift (dark gray or white). Mean ± SEM for seven participants. Trunk (FE = flexion-extension, LI = lateral inclination, TO = torsion), scapula (PR = protraction-retraction, MLR = mediolateral rotation, APT = anteroposterior tilting), glenohumeral joint (HA = horizontal abduction, E = elevation, IER = internal/external rotation), elbow (FE = flexion-extension, AA = adduction abduction, PS = prono-supination), and wrist (FE = flexion-extension, AA = adduction abduction, TO = torsion). The angles are given for each segment in the order of Euler angle rotation sequence. For the sake of clarity, the main directions of rotation are considered as positive values.
A “fixed but rotating” scapular model is obviously an oversimplification since the scapula may translate due to the sternoclavicular and acromioclavicular joints (van der Helm, 1994). The present study gives an estimate of the large 3-D translation of the scapula during reaching and demonstrates that it largely contributes to the trajectory of the hand. In addition, we show that the scapulothoracic gliding surface could be approximated by a sphere or ellipsoid (Maurel & Thalmann, 2000).

The Euler angle computation was less reliable at the elbow: it gave no physiological values of elbow abduction-adduction and propagated a non-negligible error at the level of the hand. These errors probably reflect the fact that the elbow axes of rotation are not accurately determined relative to the sensors. An oversimplification of the elbow joint probably contribute to the errors (Biryukova et al., 2000; Ishizuki, 1979; Prokopenko et al., 2001) since the ~2 cm error range is consistent with studies on elbow functional anatomy showing displacements of the instantaneous helical axis of elbow flexion-extension in vivo (Stokdijk et al., 1999) or in cadaver experiments (Veeger et al., 1997). The measure of elbow flexion-extension appeared reliable (similar to the 3-D elbow angle), but the measure of prono-supination, which is challenging to record whatever the method used, should be taken with great care.

The wrist is usually described with two degrees of freedom, when we observed around 5° rotation in the third Euler angle (torsion). This may be due to measurement error, but we cannot exclude the possibility that we captured some of the joint complexity, particularly at the end of the movement when the subject is lifting an object.

As a whole, we conclude that a rigid-body model of the trunk and upper limb with joints rotations represented by Euler angles can give reliable results propagated at the level of hand movements, providing that the translation of the scapula is taken into account. The method using electromagnetic sensors for biomechanical analysis should be further validated by estimation of the skin artifacts and other measurement errors (e.g., by comparison with other sensors). The direct kinematic model provides a virtual mannequin that could be used for other applications, for example, computer graphics (Maurel & Thalman, 2000), clinical applications, or analysis of the contribution of the proximal shoulder girdle to the arm reaching workspace adapted to individual morphology (Klopcar et al., 2007; Wang, 1999).
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References


Van der Helm, F.C. (2002). A standardized protocol for the study of shoulder motions. Standardization of joint rotation, ISG (International Shoulder Group) proposal. ISG Committee on Standardized Description of Shoulder Motion.


