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Biol. Cybern. 86, 167–177 (2002)
DOI 10.1007/s00422-001-0286-x

Biological Cybernetics

Erratum

A method for estimating torque-vector directions of shoulder muscles using surface EMGs

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Received: 17 January 2001 / Accepted in revised form: 20 September 2001

Abstract. In this study, a new method is proposed to estimate the torque-vector directions of each shoulder muscle. The method is based on a multiple regression model that reconstructs shoulder torque, which is calculated from the hand force and posture, from the surface EMG of many muscles recorded simultaneously. The torque-vector directions of eleven shoulder muscles of four subjects were obtained at up to 30 different arm postures with this method. The mean confidence interval ($p < 0.05$) of the estimated torque-vector direction of each subject was 7.7–10.6°. The correlation coefficient between the measured shoulder torque and reconstructed shoulder torque was between 0.76–0.84. The results for majority of the muscles were in accordance with previous studies, and reasonable from the viewpoint of anatomy. The torque-vector directions of a muscle, which are estimated with this method, have more of a functional meaning than a pure anatomical or mechanical one. These indicate the direction of the shoulder torque accompanying the muscle activation for a normal shoulder action that involves the cooperative contraction of many muscles.

1 Introduction

For mono-axis joints, the relationships between muscles and their action directions are clear. The muscles around such joints can easily be categorized as flexor or extensor muscles from the locations of their origins and insertions. In contrast, for multi-axis joints such as a shoulder joint, the same relationships are rather unclear. This is because a shoulder joint has three motion directions (degrees of freedom), and a number of muscles are involved in a shoulder action. Few studies have attempted to quantify the movement directions and their postural variations for many shoulder muscles.

The action directions of muscles have been obtained by means of anatomical dissection in several studies. For example, Wood et al. (1989) and Van der Helm and his colleagues (Meskers et al. 1998; Van der Helm 1994a, 1994b; Van der Helm and Veenbaas 1991; Van der Helm et al. 1992; Veeger et al. 1991) estimated the action directions of many shoulder muscles based on precise measurements of the origins, insertions, and fiber trajectories of these muscles in cadavers. Bassett et al. (1990) used a method of computer-assisted gross muscle cross-section analysis. In these methods with cadavers, however, there are the following problems. In many cases, only one posture was used. Furthermore, the estimations were based on rather arbitrary lines of muscle actions and center–of–rotation actions of the joints. These methods are indirect in estimating the action directions because real muscle forces or torques were not measured.

Buneo et al. (1997) rather directly estimated the action directions of three shoulder muscles (six parts) by electrical stimulation. Electrical stimulation was applied to the point of contraction of each muscle, and the resulting forces and torques were recorded for up to 29 different arm postures. This method was outstanding, since it measured the muscle forces and torques directly to estimate the action directions. Nevertheless, problems remained. For example, the number of muscles under study was only three and the muscles were stimulated under the condition that the subject could exert forces voluntarily or subconsciously in muscles other than those induced by electrical stimulation.

In this study, we developed a new method to estimate the action directions of muscles. We recorded hand forces, EMG, and postures, and then obtained estimations of the torque-vector directions of eleven shoulder muscles in four subjects at 12–30 different arm postures based on a multiple regression model.

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2 Muscle activation and joint torque

Torque τ of a mono-axis joint model is calculated as follows, as shown in Fig. 1:

$$\tau = LT \sin\theta \tag{1}$$

where T is the muscle tension and L is the length from the center of rotation p_1 to the point of action p_2. The product $L \sin \theta$ is called the moment arm. Generally, the magnitude of the moment arm depends on the joint angle or posture. On the other hand, for a joint with unfixed axes (such as a ball joint), not only the magnitude of the torque but also the direction of the rotation axis is needed to represent the joint torque. The torque vector that shows the absolute magnitude of the joint torque by its length and the direction of rotation axis by its orientation, is calculated as the outer product of $\boldsymbol{L}$ and $\boldsymbol{T}$:

$$\boldsymbol{\tau} = \boldsymbol{L} \times \boldsymbol{T} \tag{2}$$

where $\boldsymbol{L}$ is the vector from p_1 to p_2, and $\boldsymbol{T}$ is the vector representing the magnitude and direction of the muscle tension. Vectors $\boldsymbol{\tau}$, $\boldsymbol{L}$, and $\boldsymbol{T}$ are all three-dimensional vectors. The rotation direction shown by the torque vector is defined by the right-hand rule: if the thumb points along the vector, the fingers curl in the direction of the rotation in response to the applied torque.

When multiple muscles participate in a joint action, the net joint torque can be calculated as a sum of the torque vector of each muscle:

$$\boldsymbol{\tau} = \sum \boldsymbol{\tau}_i = \sum \boldsymbol{L}_i \times \boldsymbol{T}_i \tag{3}$$

where i is the muscle index. The torque vector of the ith muscle is written as

$$\begin{aligned} \boldsymbol{\tau}_i &= \boldsymbol{L}_i \times \boldsymbol{T}_i \\ &= |\boldsymbol{T}_i| \cdot |\boldsymbol{L}_i| (\sin\theta_i) \boldsymbol{e_i} \\ &= |\boldsymbol{T}_i| \boldsymbol{\alpha}_i \end{aligned} \tag{4}$$

where $\boldsymbol{e_i}$ is a unit vector in the same direction as $\boldsymbol{\tau}_i$; we call $\boldsymbol{\alpha}$ the "moment arm vector" here. As shown in (4), the directions of the torque vector and the corresponding moment arm vector are the same. The length of the moment arm vector is equivalent to the ratio between the muscle tension and the consequent torque. As long as the arm posture is constant, $\boldsymbol{L}$ and θ (angle of $\boldsymbol{L}$ and $\boldsymbol{T}$) are also constant. Therefore, a particular set of $\boldsymbol{\alpha}_i$ exists for a particular posture. Using an $\boldsymbol{\alpha}$ of a muscle at a given posture, we can predict the exact moving direction of the arm by an action of that muscle.

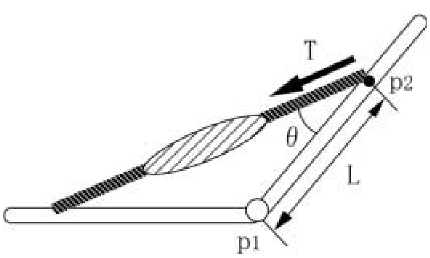

Fig. 1. Model of a mono-axis joint

3 Methods

3.1 Subjects

The subjects were four right-handed male university students (21–22 years old) with no previous history of neuropathies or trauma to the upper limbs. The subjects were given sufficient information about the experiment, and then gave their consent to participate. The institutional ethics committee approved the experiments.

3.2 Experimental design

Figure 2 shows a schematic drawing of the experimental setup. Each subject sat in a chair and had the body restrained by a harness, but the scapula was free to move. The subject gripped the handle of a transducer tightly. The wrist joint was not fixed. During each trial, the subject's arm was kept in a specific posture (12–30 different postures, see Fig. 3). The arm posture – both the shoulder and elbow joint positions – was varied by moving the transducer fixed on a table, which was movable and height-adjustable. At one transducer position, three arm postures were assigned by setting the elbow level as high, middle, or low. The number of experimental postures across the subjects differed, since the number was set considering the condition of each subject, such as the level of fatigue. Several tentative transducer positions were selected before the experiment within a comfortable reaching range. In the experiment, the transducer positions were set based on the tentative positions and adjusted so as, not to make the subjects uncomfortable. In each experiment of subjects 2 and 3, the transducer positions were at the same level. After trials with almost the same transducer position of subject 3, additional trials were carried out at higher levels of the transducer position in the experiments of subjects 1 and 4.

The subject gripped the handle of a six-degree-of-freedom force–torque transducer (UGS 3012A45, Nitta Osaka, Japan) and exerted various kinds of required forces while watching a CRT display showing the shoul-

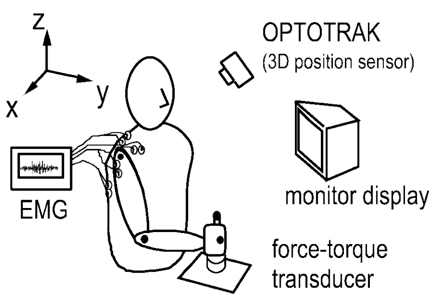

Fig. 2. Schematic drawing of the experimental setup

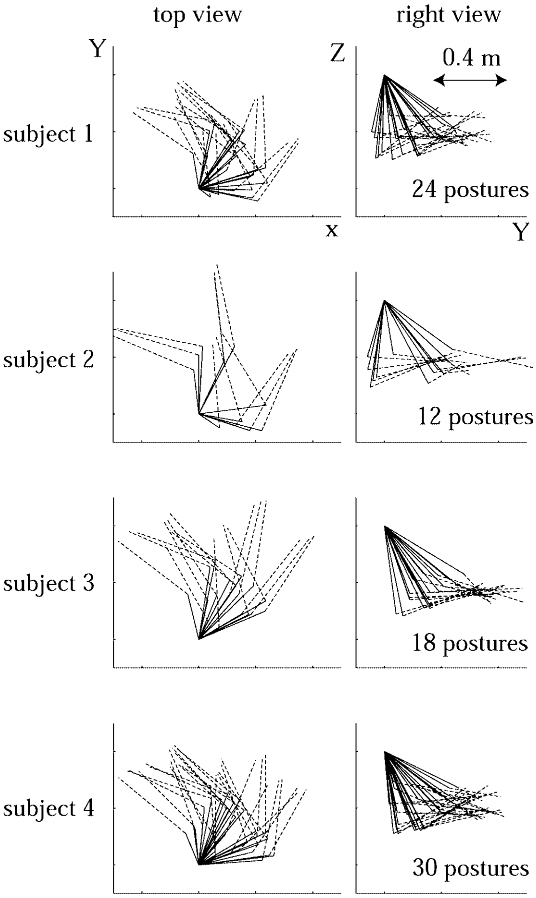

Fig. 3. Measured arm positions. The *solid lines* represent the upper arm and the *dashed lines* represent the forearm. The shoulder position is located at the origin

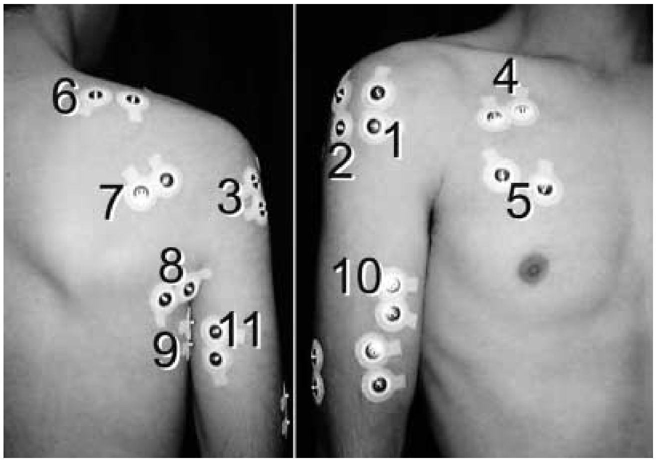

Fig. 4. Electrode positions for EMG measurement. The numbers correspond to the muscles as follows:
(1) AD anterior deltoid (clavicular part of deltoid)
(2) MD middle deltoid (acrominal part of deltoid)
(3) PD posterior deltoid (scapular part of deltoid)
(4) CP clavicular part of pectoralis major
(5) SP sternocostal part of pectoralis major
(6) T/S upper part of trapezius/supraspinatus
(7) I/T infraspinatus/teres minor
(8) TM teres major
(9) LD latissimus dorsi
(10) BB long head of biceps brachii
(11) TB long head of triceps brachii

der torque. The translation forces in the x, y, and z directions and the rotation moment on the x-, y-, and z-axes were sampled at 200 Hz with 12-bit resolution. To generate a wide range of directions and magnitudes of the shoulder torque, ten kinds of guide patterns were instructed to the subjects. They included, for example, "draw crosses on the display" or "draw whirls with an imaginary pen on the top of the handle of the transducer". No directions were given for muscle co-contraction or joint stiffness, such as "relax as much as you can" or "make your arm stiff". The subjects rehearsed how to generate forces using the experimental setup on the day before the experiment. The duration of one trial for one guide pattern was 6 s, so the total duration of ten trials for one posture was 60 s.

Three out of ten kinds of guide patterns were for translation forces and the others were for rotation moment at the handle of the transducer. The shoulder torques were calculated (see Sect. 3.3.2) in real time and shown on the display. The subjects generated hand forces and moments whilst, following the guide pattern watching the shoulder torque shown on the display. The subjects performed a single trial of each guide pattern unless the performance deviated from the pattern. In the trials to generate translation forces, the shoulder torques, which were projected on a plane perpendicular to the straight line connecting the shoulder and the hand, were shown on the display because translation forces at the hand are originated only from the component of the shoulder torque vector in parallel to the plane (see Appendix). In the trials to generate a rotation moment, the shoulder torques projected on the xy-plane and the xz-plane were shown on the display.

The positions of infrared light-emitting diode markers on the shoulder, elbow, hand, and top of the handle of the transducer were recorded at 100 Hz using a three-dimensional position sensing system (OPTOTRAK, Northern Digital, Waterloo, Ontario, Canada). When the position of each marker moved over 3 cm from the prescribed standard position for each posture, an automatic alarm was given and the data were discarded.

We used pairs of silver/silver chloride surface electrodes (1 cm in diameter) in a bipolar configuration to record the EMG signals. The signals were differentially amplified and filtered (20–1500 Hz) using a multichannel biomedical amplifier (Neurotop MME-3132, Nihon Kohden, Tokyo, Japan) and sampled at 2 kHz with 12-bit resolution. The muscles measured were as in (Fig. 4).

It is hard to record EMG signals from supraspinatus solely by surface electrodes, since this muscle is covered by the upper part of the trapezius. Accordingly, in this study we placed electrodes above the spine of the scapula (see Fig. 4, electrode 6) and treated these muscles as one functional group. Although some parts of infraspinatus and teres minor are also covered by the trapezius, it is possible to record EMG signals from these muscles by surface electrodes located near the insertions of the muscles (see Fig. 4, electrode 7). It is, however, not easy to distinguish infraspinatus and teres minor near these insertions, since the insertions are very close to each other and the muscle fibers of both run in almost the same direction. Therefore, we treated infraspinatus and teres minor as a functional group in this study.

In addition to these 11 muscles, we recorded EMG of the lateral head of triceps brachii and brachialis. Because these muscles are not concerned with shoulder motion directly, we did not use the data from these muscles in this study. The short head of biceps brachii was not measured because of a limitation of the particular channel of the EMG recording system.

3.3 Analysis

3.3.1 EMG data analysis. EMG signals were rectified and then filtered (50-point moving-average window) digitally. After that, the signals were smoothed with the finite impulse response filter proposed by Koike and Kawato (1995). The filter had an impulse response as follows:

$$h(t) = 6.44(\exp(-10.80t) - \exp(-16.52t)) \quad (5)$$

where t is the time. This smoothing method was developed in a study (Koike and Kawato 1995) into the tension of shoulder muscles under isometric contraction. The EMG signals smoothed with this method were regarded as almost proportional to the muscle tension. Accordingly, the magnitude of $\boldsymbol{T}$ can be shown as

$$|\boldsymbol{T}| = kE \quad (6)$$

where k is a coefficient of proportion and E is the smoothed EMG.

3.3.2 Shoulder torque. The shoulder torque is calculated as follows (see the Appendix):

$$\boldsymbol{\tau} = \boldsymbol{L}_\mathrm{h} \times \boldsymbol{F} + \boldsymbol{M} - \boldsymbol{L}_\mathrm{a} \times (W_\mathrm{a}\mathbf{g}) \quad (7)$$

where $\boldsymbol{L}_\mathrm{h}$ is a vector from the shoulder to the hand, $\boldsymbol{F}$ is the translation force of the hand, $\boldsymbol{M}$ is the moment of the hand, $\boldsymbol{L}_\mathrm{a}$ is a vector from the shoulder to the center of mass of the upper limb, W_a is the mass of the upper limb, and $\boldsymbol{g}$ is the acceleration of gravity. From the regression of an equation of Zatsiorsky et al. (1990), $\boldsymbol{L}_\mathrm{a}$ and W_a were estimated from the weight and height of the subject.

In the study of Buneo et al. (1997), shoulder torques were defined in both arm-fixed and body-fixed frames of reference. The frame of reference used in this study was body fixed.

3.3.3 Estimation of torque-vector direction. The shoulder torque in (7) can be represented as follows based on (3), (4), and (6):

$$\begin{aligned}\boldsymbol{\tau} &= \sum \boldsymbol{\alpha}_i|\boldsymbol{T}_i| + \boldsymbol{b}\\ &= \sum \boldsymbol{\alpha}_i kE_i + \boldsymbol{b}\\ &= \sum \boldsymbol{a}_i E_i + \boldsymbol{b}\end{aligned} \quad (8)$$

where $\boldsymbol{a} = k\boldsymbol{\alpha}$ and $\boldsymbol{b}$ is a bias term related to the tension of unrecorded muscles and/or other tissues. Equation (8) is formulated as a multiple regression model. Therefore, $\boldsymbol{a}_i$ can be estimated from $\boldsymbol{\tau}$ and E_i by means of multiple regression analysis. As a result, the torque-vector direction of each muscle can be estimated, since the direction of $\boldsymbol{a}_i$ is equivalent to the torque-vector direction of the $\boldsymbol{\alpha}_i$ values of the ith muscle. In this study, 12 000 points (60 s × 200 Hz) of τ were calculated for each posture. We used every other point of $\boldsymbol{\tau}$ (6 000 points) and the corresponding E_i values to estimate the $\boldsymbol{a}_i$ values of 11 muscles at each posture.

4 Results

4.1 Reliability of estimated torque vectors

In this study, we focused only on the direction of vector $\boldsymbol{a}$; its length was not systematically explored. Therefore, we represented the confidence interval of only its direction in degrees. The elements of x, y, and z of vector $\boldsymbol{a} = [a_x, a_y, a_z]^\mathrm{T}$ were estimated and the confidence interval of each element, S_x, S_y, and S_z were defined under a specified level of significance:

$$\begin{aligned}a_x - \Delta a_x &\le S_x \le a_x + \Delta a_x\\ a_y - \Delta a_y &\le S_y \le a_y + \Delta a_y\\ a_z - \Delta a_z &\le S_z \le a_z + \Delta a_z\end{aligned} \quad (9)$$

The vector set $[S_x, S_y, S_z]^\mathrm{T}$ makes the rectangular prism with vertexes $\boldsymbol{a}' = [a_x \pm \Delta a_x, a_y \pm \Delta a_y, a_z \pm \Delta a_z]^\mathrm{T}$. We defined the confidence interval of the angle of $\boldsymbol{a}$ to be the largest angle among all angles between $\boldsymbol{a}$ and $\boldsymbol{a}'$:

$$C = \max\{A(\boldsymbol{\alpha}, \boldsymbol{\alpha}')\} \quad (10)$$

where C is the confidence interval of the angle of $\boldsymbol{a}$, and A is the function to calculate the angle between two vectors. The mean of the confidence angel of $\boldsymbol{a}$ of each subject was defined as follows:

$$\left(\sum\sum C_{i,j}\right)/(N_m \cdot N_p) \quad (11)$$

where i is the index of muscle, j is the index of experimental arm posture, N_m is the number of muscles, and N_p is the number of arm postures. At $p < 0.05$, the means of the confidence angles of $\boldsymbol{a}$ were 8.5°, 10.6°, 7.7°, and 9.2° for the subjects.

4.2 Reconstruction of shoulder torque

Shoulder torques can be reconstructed from torque vectors and smoothed EMGs using (8). To evaluate the reliability of our estimated torque vectors, the experimental data were divided in half and the shoulder torque in one half of the data were reconstructed using the estimated torque vectors from the other half of the data. As shown in Sect. 3.2, there were ten trials for one posture corresponding to ten kinds of guide patterns of

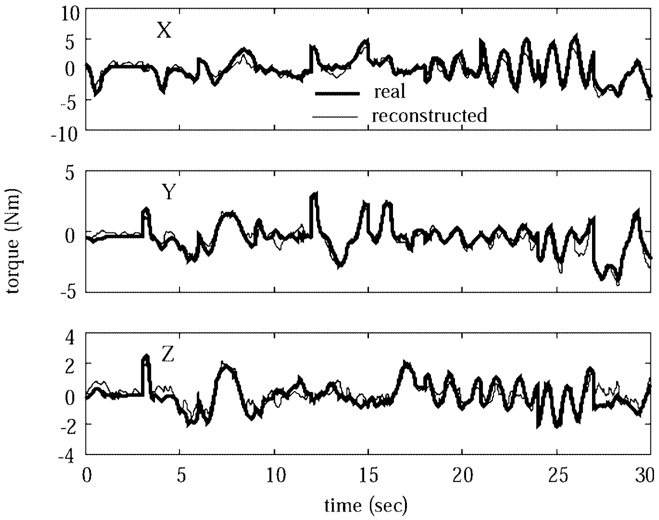

Fig. 5. An example of reconstructed and real shoulder torques

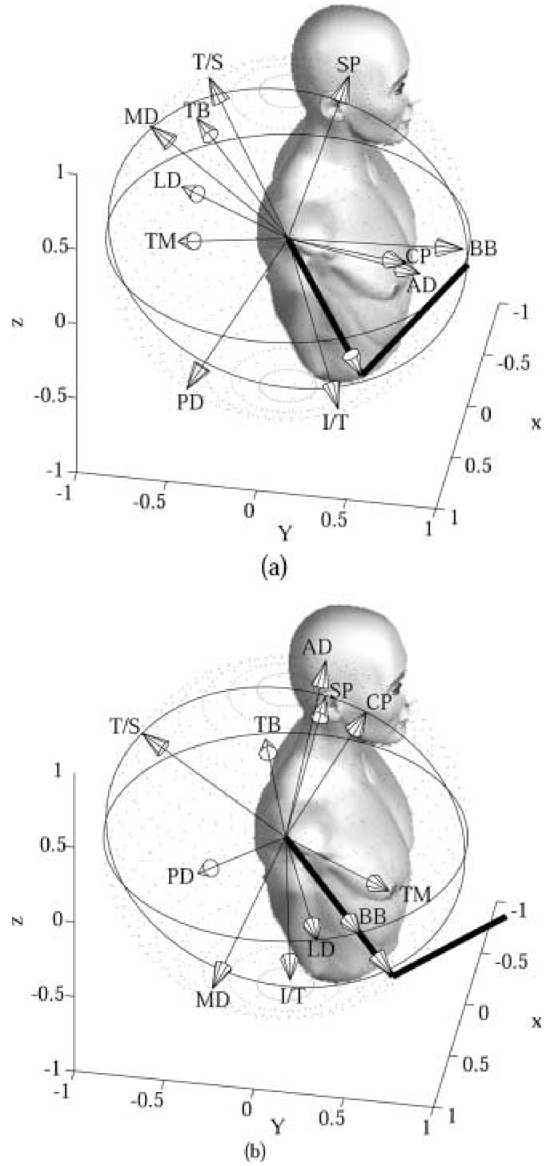

Fig. 6. Torque-vector directions of 11 muscles for two example arm positions. The *thick lines* present directions of the arm

the shoulder torque. One trial took 6 s. First, the torque vectors were estimated using the first 3 s of the data from each trial. Then the shoulder torque was reconstructed using the last 3s from the estimated torque vectors and the EMG, using (8). The correlation between the reconstructed shoulder torque and the real one was evaluated. The means of the correlation coefficients of the torques in the x-, y- and z-axes for every posture of the four subjects were 0.84, 0.76, 0.82, and 0.76.

Figure 5 shows an example of a 30-s record from a reconstructed shoulder torque. The reconstructed torque is very close to the real one.

4.3 Analysis results for each posture

Figure 6 shows examples of the torque-vector directions of eleven muscles for two postures of the same subject. The start points of the vectors are at the origin and their lengths are normalized to unity. The thick lines in the figure show the directions of the upper arm and forearm. The direction of each torque vector is given by the right-hand rule. For example, the torque vector of infraspinatus/teres minor for the posture in Fig. 6a is close to the direction of the longitudinal axis of the upper arm. This means that the muscle acted in an external rotation of the shoulder. The anterior deltoid and the clavicular part of pectoralis major – muscles that are assumed to have similar functions from the viewpoints of origin and insertion—were estimated to have similar torque-vector directions for both postures. The torque-vector directions of other muscles were also reasonable from the viewpoint of anatomy. Numerically represented torque-vector directions of each muscle for some example postures of subject 1 are tabulated in Table 1.

4.4 Results of an analysis of each muscle

Figure 7 shows the torque-vector direction of each muscle for every posture. The movement directions to which each muscle contributes and their variations over different postures can be understood from this figure.

The range of vector directions shows that some pairs of muscles are in an agonist–antagonist relationship, such as anterior and posterior deltoid (muscles 1 and 3), the sternocostal part of pectoralis major and infraspinatus/teres minor (muscles 5 and 7), and biceps and triceps (muscles 10 and 11). These are reasonable results from the viewpoint of anatomy. The variations in the vector direction show that some muscles do change their torque-vector directions widely depending on the posture, whereas others do not. Let us view the anterior deltoid (muscle 1) and clavicular part of pectoralis major (muscle 4) as an example. Anatomy suggests that these muscles should have similar torque-vector directions. In fact, the ranges of the estimated torque-vector directions

Table 1. Torque-vector directions of each muscle of subject 1 for example postures

Posture no.			1	2	3	4	5	6	7	8	9	10	11	12	13
	Upper arm direction	x	0.14	0.72	0.63	−0.07	0.19	0.12	0.13	0.55	0.43	0.41	0.71	0.64	0.33
		y	−0.10	0.23	0.14	0.40	0.78	0.66	0.09	0.52	0.37	−0.03	−0.15	−0.12	0.21
		z	−0.99	−0.66	−0.77	−0.91	−0.60	−0.74	−0.99	−0.66	−0.82	−0.91	−0.69	−0.76	−0.92
	Forearm direction	x	−0.10	−0.64	−0.56	−0.77	−0.95	−0.95	−0.46	−0.83	−0.77	0.71	0.57	0.61	0.35
		y	0.90	0.74	0.83	0.53	0.18	0.30	0.81	0.52	0.64	0.63	0.82	0.79	0.89
		z	0.43	−0.20	0.01	0.37	−0.26	−0.05	0.37	−0.20	0.07	0.31	−0.10	0.02	0.29
1	Anterior deltoid	x	0.58	−0.22	0.10	0.52	0.85	0.58	0.82	0.96	0.93	0.46	0.31	0.44	0.92
		y	−0.06	0.23	0.27	0.83	0.53	0.66	0.42	0.06	0.23	0.18	−0.16	−0.31	−0.17
		z	0.81	0.95	0.96	0.22	−0.01	0.47	−0.38	0.28	0.27	0.87	0.94	0.84	0.36
2	Middle deltoid	x	−0.93	0.56	0.55	−0.38	0.95	0.64	−0.18	0.66	−0.04	0.56	−0.02	−0.54	0.39
		y	0.34	−0.67	−0.67	−0.84	−0.02	−0.69	−0.38	−0.74	−1.00	−0.83	−0.99	−0.84	−0.70
		z	−0.15	−0.49	−0.51	0.39	−0.32	−0.33	0.91	−0.14	−0.09	0.00	−0.11	0.06	0.61
3	Posterior deltoid	x	−0.86	−0.80	−0.53	0.49	0.16	0.34	−0.87	−0.50	−0.54	−0.98	−0.46	−0.82	−0.99
		y	−0.43	−0.34	−0.71	−0.48	−0.56	−0.36	−0.49	−0.63	−0.76	−0.18	−0.19	−0.18	−0.09
		z	0.28	−0.49	−0.47	−0.73	−0.82	−0.87	−0.04	−0.59	−0.36	0.00	−0.87	−0.55	0.05
4	Clavicular part of pectoralis major	x	0.64	−0.04	0.37	0.57	0.17	0.26	0.54	−0.30	0.33	0.11	0.01	0.29	0.42
		y	0.62	0.70	0.68	0.76	0.83	0.68	0.73	0.63	0.76	0.78	0.80	0.50	0.41
		z	0.45	0.71	0.63	0.32	0.53	0.69	0.42	0.72	0.57	0.61	0.60	0.82	0.81
5	Sternocostal part of pectoralis major	x	−0.08	−0.64	−0.76	−0.33	−0.76	−0.88	−0.19	−0.46	−0.69	−0.07	−0.31	−0.26	−0.39
		y	0.57	0.63	0.42	0.28	0.03	0.02	0.55	0.57	0.33	0.80	0.90	0.86	0.68
		z	0.82	0.45	0.50	0.90	0.65	0.48	0.81	0.68	0.65	0.59	0.31	0.43	0.62
6	Upper trapezius/ supraspinatus	x	0.32	−0.42	−0.94	−0.30	−0.49	−0.30	−0.72	−0.87	−0.60	0.11	−0.66	−0.38	−0.54
		y	−0.66	0.17	0.03	−0.49	−0.74	−0.87	0.20	−0.28	−0.08	0.86	0.31	0.79	0.82
		z	−0.68	0.89	0.35	0.82	0.46	0.40	0.66	0.42	0.79	0.50	0.68	0.49	0.18
7	Infraspinatus/teres minor	x	0.71	0.96	0.92	0.65	0.74	0.73	0.80	0.92	0.90	0.41	0.39	0.48	0.56
		y	−0.36	−0.11	−0.18	0.40	0.59	0.58	−0.06	0.38	0.13	−0.71	−0.83	−0.77	−0.50
		z	−0.60	−0.25	−0.36	−0.65	−0.33	−0.35	−0.60	−0.12	−0.42	−0.58	−0.39	−0.42	−0.65
8	Teres major	x	−0.97	−0.85	−0.98	−0.53	−0.03	0.32	−0.78	0.21	−0.97	−0.64	−0.56	−0.55	−0.43
		y	−0.04	0.38	−0.12	−0.71	−0.36	0.66	−0.62	−0.77	−0.22	0.76	0.83	0.83	0.69
		z	0.25	0.36	−0.17	−0.47	0.93	−0.68	−0.11	−0.60	0.10	0.14	0.05	0.11	−0.58
9	Latissimus dorsi	x	−0.78	1.00	0.98	−0.67	0.65	−0.51	−0.90	0.84	0.91	0.61	0.83	0.77	0.97
		y	−0.41	0.07	0.18	−0.71	0.72	−0.57	−0.24	0.14	0.38	−0.78	−0.55	−0.64	−0.20
		z	−0.47	−0.01	−0.08	−0.20	0.23	0.64	−0.37	−0.52	−0.18	−0.11	0.05	−0.07	−0.13
10	Biceps brachii (long head)	x	0.83	0.91	0.88	0.07	0.55	0.33	−0.11	0.80	0.08	0.97	0.96	0.92	0.97
		y	0.39	0.29	0.21	0.99	0.83	0.94	0.94	0.59	0.96	−0.11	−0.22	−0.37	0.12
		z	−0.40	−0.31	−0.42	0.10	0.02	0.06	−0.33	−0.06	−0.27	0.19	0.16	0.07	−0.22
11	Triceps brachii (long head)	x	−0.68	−0.63	−0.89	−0.74	−0.73	−0.32	−0.95	−0.80	−0.89	−0.64	−0.80	−0.81	−0.83
		y	0.67	−0.58	0.16	−0.64	−0.67	−0.91	−0.20	−0.60	−0.44	0.76	0.50	0.57	0.45
		z	0.29	0.51	0.42	0.22	−0.09	−0.25	0.24	−0.07	−0.06	0.11	−0.34	−0.13	0.34

of both muscles overlapped. The variations of each muscle, however, differed.

The postural dependence of the torque-vector direction can be seen from the data acquired with this method. For example, let us focus on the long head of biceps brachii, which showed a rather simple postural dependence. The long head of biceps brachii arises from the supraglenoid tubercle, and its long tendon descends in the intertubercular groove. The tendon and the groove can be likened to a rope and pulley system. Therefore, it was assumed that the torque-vector directions of the muscle would vary depending on the posture of the humerus. Figure 8 shows the relationship between the azimuth of the upper arm and the azimuth of the torque vector. Here, the azimuth was defined as the angle that the projection of the upper arm or torque vector onto the horizontal plane would make with the x-axis; in other words, the azimuth was a rotation angle about the z-axis (vertical axis). The azimuth of the upper arm and that of the torque vector show clear correlations for all subjects except subject 3.

5 Discussion

5.1 Comparisons with previous studies

We have shown that the action directions of many shoulder muscles can be estimated quantitatively using the method that we applied in this study. We compared the results obtained by the method with data obtained in previous studies.

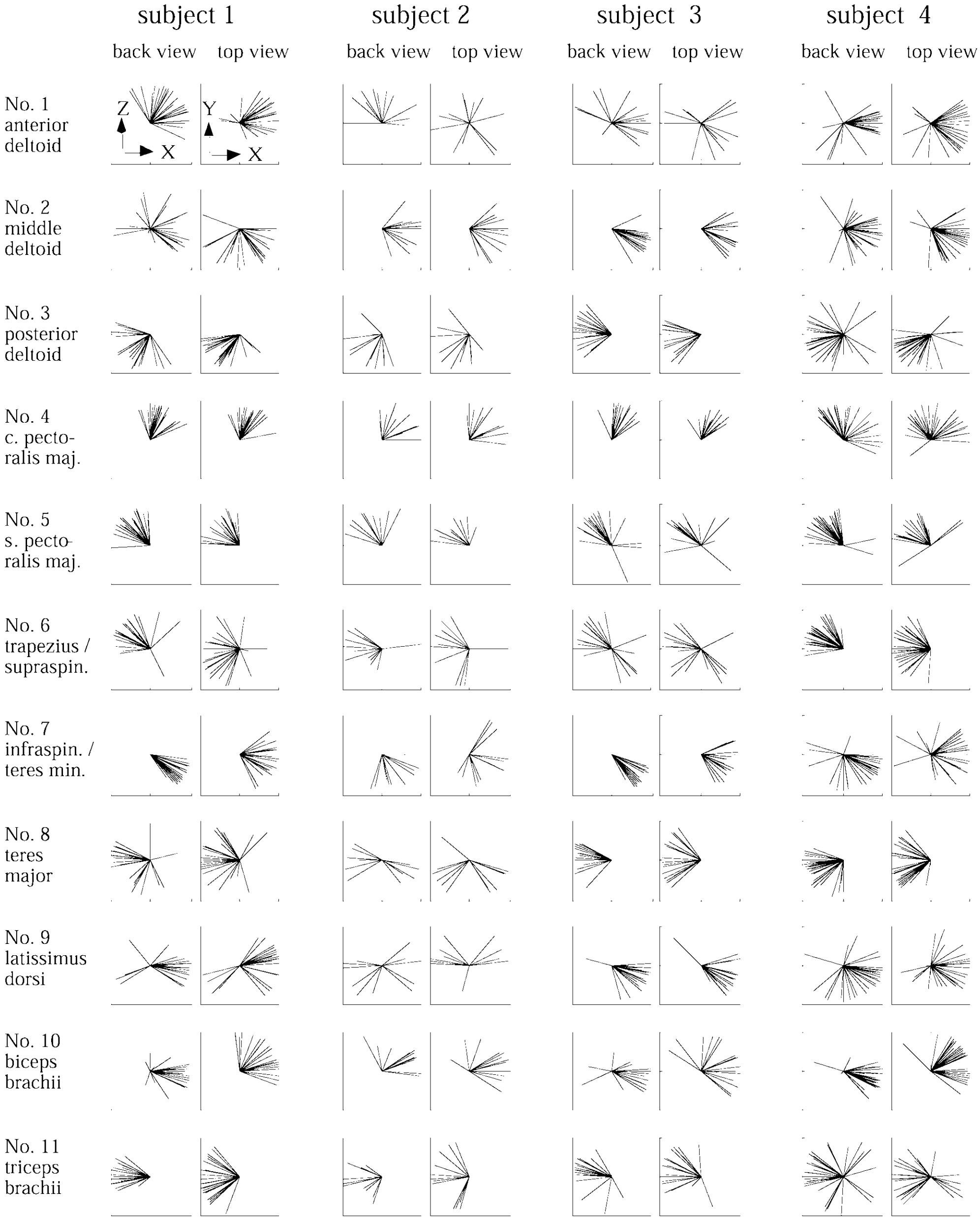

Fig. 7. Torque-vector directions of each muscle for every arm position. The numbers correspond to the muscles as in Fig. 4

Wood et al. (1989) estimated the moment arms of many shoulder muscles based on several pairs of origins and insertions for muscles that were measured in detail, but their data related only to a single arm posture of one cadaver. Torque-vector directions calculated from this data are shown in Fig. 9 (upper graph). To compare these data with the results of the present study, the posture of each subject that was nearest to the cadaver's posture was selected. The nearness of the two postures was evaluated using the angle of rotation of the upper arm, which transforms one posture to another one. The angle of rotation uses not only the long axis direction of the upper arm, but also the humeral rotation direction. The angles between the cadaver's posture and the nearest posture of each subject were evaluated to be 5.4°, 25.2°, 14.9°, and 14.6°. The lower graph of Fig. 9 shows the torque-vector directions of muscles for the first subject whose posture was nearest to that of the cadaver. In particular, the torque-vector directions of muscles of anterior deltoid, middle deltoid, posterior deltoid, and the clavicular part of pectoralis major in the two graphs were very close (< 20°difference). The differences in the torque vectors for all of the muscles between the cadaver and all subjects in the nearest posture were calculated. The mean differences in all of the muscles of all subjects was 66.5°. Every mean difference of each muscle except supraspinatus (119.3°), latissimus dorsi (145.5°) and teres minor (138°) was smaller than 70° for all of the subjects.

Buneo et al. (1997) constructed a regression model that estimates the torque-vector direction from the shoulder posture, based on torque-vector direction data obtained by electrically stimulating muscles for up to 29 different arm postures. Figure 10 shows torque-vector

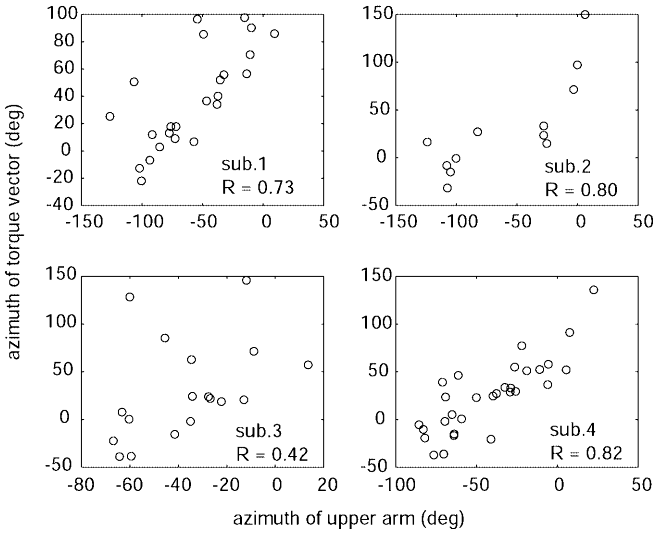

Fig. 8. Scatter graph of the azimuth of the upper arm and the azimuth of the torque vector (long head of biceps brachii)

directions calculated by the model of Buneo et al. for the posture data of subject 1 obtained in this study. In this study, the range of shoulder postures was similar for all subjects; therefore, torque-vector directions calculated by Buneo's model from the postural data were in almost the same range for different subjects. In Fig. 10, there is only one vector in the top-view graph of muscles 1 (anterior deltoid) and 3 (posterior deltoid), because the horizontal component of the torque-vector directions of these muscles was regarded as independent of the posture in the regression model of Buneo et al. In general, the torque-vector directions obtained in this study agree with the results estimated by Buneo's model as shown in Fig. 7 and 10, except the result of latissimus dorsi (muscle 9) which showed a discrepancy of about 90°.

To compare the results estimated by Buneo's method with the results of this study, some methodological differences should be taken into consideration. One of these differences is in the electrode placement. The position, shape, size, and interval of the electrodes might have affected the results in both methods. Additionally, the elbow angle was fixed at 90° flexion in Buneo's method, but this was varied in our method. Since the shoulder muscle group includes biarticular muscles which cross the elbow joint as well as the shoulder joint, the angle variation may also influence the results.

5.2 Advantages and disadvantages of each estimation method

In this paper, we described the possibility of estimating the torque-vector directions of shoulder muscles by utilizing EMG–torque relationships. The directions obtained with this method agreed, in general, with results from an anatomical method and an electrical stimulation method. In the following, we discuss the advantages and disadvantages of each method.

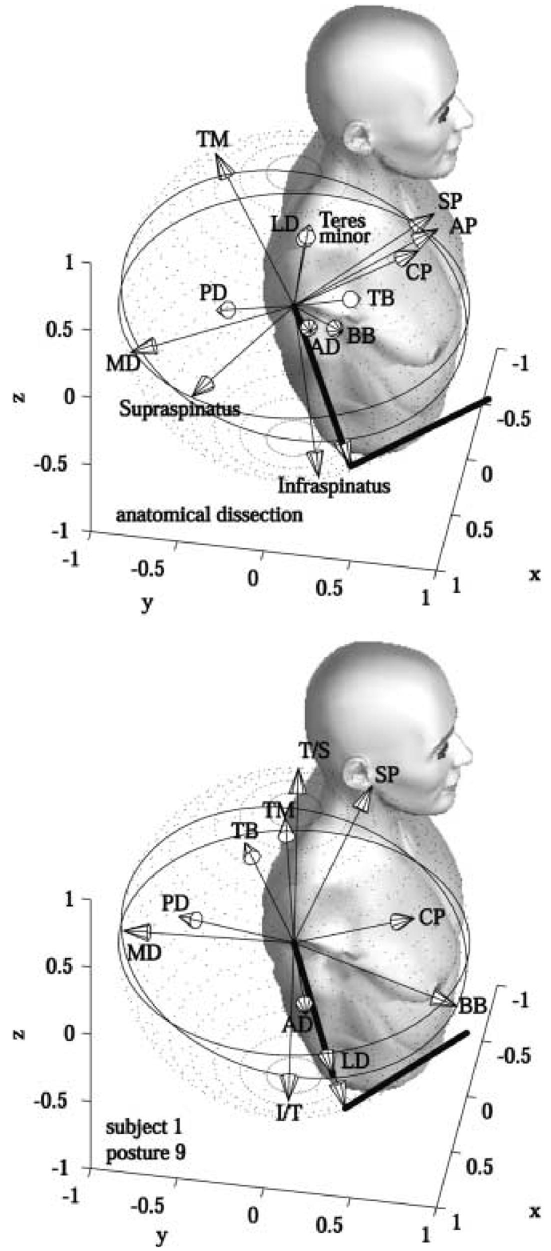

Fig. 9. Torque-vector directions estimated via anatomical dissection (*upper graph*) and results of this study at almost the same posture (*lower graph*). The *thick line* presents the direction of the arm

Methods using cadavers have been the standard when estimating the functions of muscles. Methods of this type, however, have several disadvantages. First, the direction of muscle tension is estimated from the direction of muscle fibers. It would be easy to estimate the direction of muscle tension with this method, when the courses of muscle fibers are almost straight and parallel. In many cases, however, the courses of muscle fibers are neither straight nor parallel. In addition, since cadaver muscles are not flexible, it is very hard to change its posture in a wide range. Even if cadavers were to take a different posture, cadaver muscles do not change their shapes as do living muscles, since cadaver muscles never

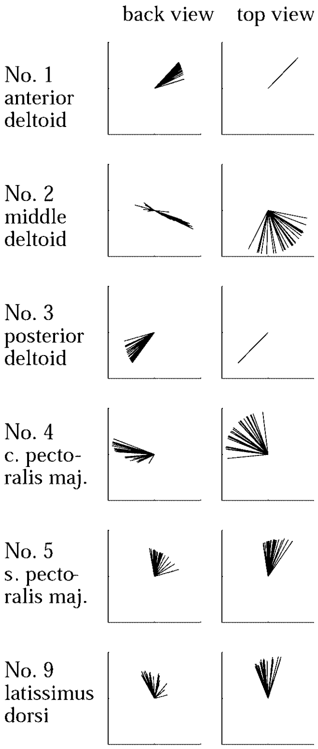

Fig. 10. Torque-vector directions calculated from the postural data of subject 1 based on a model made from an electrical stimulation study

contract. Therefore, morphological data can be obtained in only one posture for one cadaver. Needless to say, this method is not available for living subjects. On the other hand, the following points should be regarded as advantages of this method. It is easier to identify muscles in the cadavers than in living subject, and both surface and deep muscle can be studied.

Electrical stimulation methods have the following advantages over methods with cadavers. First, joint torques resulting from muscle contraction can be measured directly with a force–torque sensor in these methods, while muscle forces need to be reconstructed from morphological data in the methods with cadavers. Second, experiments can be performed for many different postures of one subject. However, there are also several disadvantages. First, muscles need to generate sufficient joint torques during the electrical stimulation. Deep muscles do not contract without the contraction of surface muscles when stimulated using a surface electrode. Moreover, even for surface muscles it is difficult to obtain sufficient torques from small muscles. Actually, the muscles that Buneo et al. (1997) used were very strong muscles, such as deltoid, pectoralis major, and latissimus dorsi. Second, it is very difficult to deny that the joint torques measured in these methods contain the torques of the muscles that were not the target of electrical stimulation. These unselected muscles might reflexively contract because of the contraction of the targeted muscles or due to the electrical stimulation itself. Furthermore, the subject might contract muscles voluntarily or unconsciously when responding to the stimulation.

The advantages of electrical stimulation methods over methods with cadavers are also advantages of our method. These include the direct measurement of muscle torques and the possibility of experimentation in many different postures. The advantages of our method over the electrical stimulation methods are as follows. First, our method is adaptable even for small muscles that are unable to generate sufficient torques by electrical stimulation, but only if EMGs of the muscles are obtained. Second, reflexive or unconscious muscle contractions do not need to be restrained. These extra contractions do not need to be taken into account explicitly, since all kinds of muscle activity appear in EMGs and they are used to estimate torque-vector directions. On the other hand, the disadvantages of our method are as follows. Using this method with surface electrodes is not very efficient for deep muscles. In our method, the EMGs of an adequate number of muscles need to be dealt with when shoulder torques have to be examined; otherwise, the estimation accuracy is unsatisfactory. In this study, all important muscles are covered except the subscapularis muscle.

5.3 Significance of the torque-vector directions obtained by the proposed method

The torque-vector directions obtained by electrical stimulation methods and methods with cadavers show the motion direction of the arm when the examined muscle alone contracts. On the other hand, the meaning of the torque-vector direction by the method proposed here is not completely the same as those methods. For example, although infraspinatus, teres minor, and subscapularis are the primarily rotators, they act during shoulder abduction and/or flexion to protect shoulder joints (Basmajian and Deluca 1985; Kapandji 1982). In our methods, thus, the motion directions of these muscles can be estimated as abduction or flexion during these actions since the activity of these muscles correlates to such actions.

When a muscle torque direction and a joint torque direction are the same, the role of the muscle can be regarded as an agonist. On the other hand, what kind of role does a muscle play when the muscle torque direction is different from the joint torque direction? One of the important roles must be the protection of joint constructions, such as that played by the rotator cuff muscles. In other words, this is a role of generating translation forces that press the humeral head to the glenoid cavity, rather than rotation forces. Adjusting the joint stiffness is also an important role. The joint stiffness can be controlled by the co-contraction level of the agonistic muscles, and the muscles having antagonistic torque components. For a multi-axis joint, the desired joint torque direction is not always generated by one muscle. In fact, the co-activation of muscles plays a role in generating the desired joint torque direction when necessary. In the torque-direction estimation method we propose in this study, if the main role of a muscle in a natural movement is not the agonistic function, the estimated torque-vector direction of the muscle is possibly different from the mechanical action direction of the muscle. That is to say, the torque vectors obtained using our method have more of a functional meaning rather than a pure anatomical or mechanical one. The different results for latissimus dorsi between our methods and others might be related to this point.

There are, however, no muscles for which the estimated torque-vector directions was opposite to their anatomical direction in this study. This does not necessarily mean that all muscles usually contribute in a positive manner. In this experiment, the subjects were asked to generate a wide range of shoulder torque directions, so every muscle might have situations in which they acted as an agonist. On the contrary, in daily living, not all muscles may act as an agonist. For example, muscles need to generate flexion torque to elevate the arm, but do not need to generate extension torque to move the arm down because gravity helps in this kind of action. Therefore, if the muscle action patterns had been limited in the experiment to those encountered in daily living, the estimated torque-vector directions would have been different from results presented here. The functional role of each muscle could be clarified by comparing results in our method and results from anatomical studies.

5.4 Applications and further investigations

The results obtained in this study might be applicable to shoulder control by functional electrical stimulation (FES), which is a rehabilitation method using electrical stimulation to restore the motor functions of people with spinal cord injury, cerebral stroke, head injury, or other disorders of the central nervous system. The relationship between a stimulated muscle and the direction of the consequent torque is clear on a mono-axis joint. However, there is a lack of quantitative data to know the appropriate combinations of stimulated muscles and stimulation magnitudes to exert desired net torques of the shoulder. The results in this study are helpful for creating stimulation patterns for shoulder FES. The shoulder joint is a ball-and-socket joint with a shallow socket. Therefore, the range of allowable motion is wide, but the construction is rather unstable. Muscles, tendons, and ligaments around the joint help to keep the ball in the socket. Accordingly, an inappropriate balance of muscle contraction might be a cause of joint disorder. To avoid this, we need to study the normal contraction balance of shoulder muscles. Hoshimiya et al. (1989) tried to use template stimulation patterns related to the EMGs of healthy persons for FES. The torque vectors obtained in this study are based on the normal balance of muscle contraction. Therefore, the torque vectors revealed here might be useful for FES.

As above mentioned, one of the important advantages of this torque-vector estimation method over anatomical methods with cadavers is the possibility of examining the postural dependence of the torque vectors. In this work, we showed a range of vector directions but elucidated only the postural dependence of the long head of biceps brachii as an example, since the prime object of this work is to prove the validity of this newly designed method. To put the torque-vector directions into practical use, in FES for example, the directions represented as a function of arm posture might be useful. Actually, Buneo et al. (1997) represented the directions estimated by a electrical stimulation method as regression models of shoulder angles shown in azimuth, elevation, and rotation. In future investigations, we intends studying the postural dependence of the torque-vector directions. An important problem in this study would be the selection of an adequate representation of the arm posture. In Table 1, we used direction vectors of upper arm and forearm to show the arm posture. There are other several kinds of representation of posture for a joint with three axes, for example Euler angles, quaternions, and rotation matrixes. Furthermore, Euler angles are subdivided into types according to the rotation sequence of the axes. A model that gives torque-vector directions from arm postures would strongly depend on the method of postural representation. Therefore, we are going to investigate the postural dependence of torque-vector directions with the method used to represent the arm posture.

Acknowledgements. We thank Mr. Kizuki Horikawa (CSK Corporation) for his help in the programming and experiments. The experiments complied with the laws of Japan. This study was performed through HFSP (OW) and Special Coordination Funds for Promoting Science and Technology.

Appendix: Method of calculating shoulder torques

As shown in Fig. A1, let the position vector of the shoulder be $\boldsymbol{S}$ and that of the center of the hand be $\boldsymbol{H}$. When the subject grips the handle of the sensor and applies a force, let the discrete contact points of the hand and the handle be $\boldsymbol{P}_1, \boldsymbol{P}_2, \cdots, \boldsymbol{P}_n$ and let the forces at the points be $\boldsymbol{F}_1, \boldsymbol{F}_2, \cdots, \boldsymbol{F}_n$. Under static conditions, the shoulder torque $\boldsymbol{\tau}_\mathrm{s}$ is calculated as follows:

$$
\begin{aligned}
\boldsymbol{\tau}_\mathrm{s} &= \sum(\boldsymbol{P}_i - \boldsymbol{S}) \times \boldsymbol{F}_i \\
&= \sum(\boldsymbol{P}_i - \boldsymbol{H} + \boldsymbol{H} - \boldsymbol{S}) \times \boldsymbol{F}_i \\
&= (\boldsymbol{H} - \boldsymbol{S}) \times \left(\sum \boldsymbol{F}_i\right) + \sum(\boldsymbol{P}_i - \boldsymbol{H}) \times \boldsymbol{F}_i
\end{aligned}
$$

where $\boldsymbol{L}_\mathrm{h}$ is used for $(\boldsymbol{H} - \boldsymbol{S})$, $\boldsymbol{F}$ for $\Sigma \boldsymbol{F}_i$, and $\boldsymbol{M}$ for $\Sigma(\boldsymbol{P}_i - \boldsymbol{H}) \times \boldsymbol{F}_i$, which shows the torque on $\boldsymbol{H}$. $\boldsymbol{\tau}_\mathrm{s}$ is:

$$\boldsymbol{\tau}_\mathrm{s} = \boldsymbol{L}_\mathrm{h} \times \boldsymbol{F} + \boldsymbol{M}$$

$\boldsymbol{\tau}_\mathrm{s}$ is the composition of muscle torque $\boldsymbol{\tau}$ and gravity torque $\boldsymbol{L}_\mathrm{a} \times (W_\mathrm{a}\boldsymbol{g})$:

$$\boldsymbol{\tau} + \boldsymbol{L}_\mathrm{a} \times (W_\mathrm{a}\boldsymbol{g}) = \boldsymbol{L}_\mathrm{h} \times \boldsymbol{F} + \boldsymbol{M}$$

Equation (6) is a transformation of the above equation.

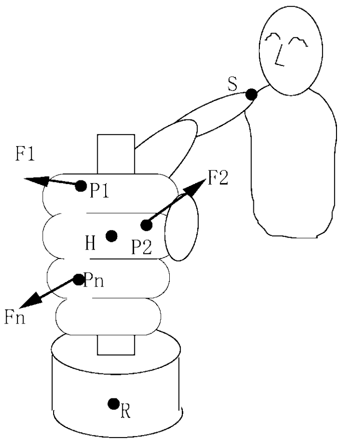

Fig. A1. Forces at the hand

In general, the center of the sensor for torque measurement, $\boldsymbol{R}$, is not at $\boldsymbol{H}$. Practically speaking, τ_s is:

$$\begin{aligned}\tau_s &= \sum(\boldsymbol{P}_i - \boldsymbol{S}) \times \boldsymbol{F}_i \\ &= \sum(\boldsymbol{P}_i - \boldsymbol{R} + \boldsymbol{R} - \boldsymbol{S}) \times \boldsymbol{F}_i \\ &= (\boldsymbol{R} - \boldsymbol{S}) \times \left(\sum \boldsymbol{F}_i\right) + \sum(\boldsymbol{P}_i - \boldsymbol{R}) \times \boldsymbol{F}_i \\ &= \boldsymbol{L}_r \times \boldsymbol{F} + \boldsymbol{M}_r\end{aligned}$$

where $\boldsymbol{L}_r$ is the vector from the shoulder to $\boldsymbol{R}$ and $\boldsymbol{M}_r$ is the torque measured with the sensor.

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