Human Arm Posture Control Using the Impedance Controllability of the Musculo-Skeletal System Against the Alteration of the Environments

Jaehyo Kim, Makoto Sato, and Yasuharu Koike

Abstract We show that humans execute the postural control ingeniously by regulating the impedance properties of the musculo-skeletal system as the motor command against the alteration of the environment. Adjusting muscle activity can control the impedance properties of the musculo-skeletal system. To quantify the changes in human arm viscoelasticity on the vertical plane during interaction with the environment, we asked our subject to hold an object. By utilizing surface electromyographic(EMG) studies, we determined a relationship between the perturbation and a time-varying muscle co-activation. Our study showed when the subject lifts the object by himself the muscle stiffness increases while the torque remains the same just before the lift-off. These results suggest that the central nervous system(CNS) simultaneously controls not only the equilibrium point(EP) and the torque, but also the muscle stiffness as the motor command in posture control during the contact task.

Keywords: electromyographic(EMG), central nervous system(CNS), muscle coactivation, equilibrium point(EP)

I. Introductions

The musculo-skeletal system is regarded as the mechanical engine driving a skeletal system, and is driven and controlled by the central nervous system. The impedance property of the musculo-skeletal system can be controlled by adjusting muscle activity. It has been pointed out the human arm viscoelasticity plays an important role in motor behavior such as voluntary movement and posture control[1][2][3][4]. The basic principle at issue in these studies is that the CNS executes the feed-forward control mechanism using the spring-like property of musculo-skeletal system. The representative approaches in these studies are the equilibrium point (EP) control hypothesis and minimum torque change model. The EP control hypothesis suggest that the CNS may execute a posture control by generating motor command which change the relative activation of the agonist and antagonist muscles. The point about the EP control hypothesis is that the muscle activation and forces depend on the difference between the arm's equilibrium position and its actual position, then the CNS controls by shifting the equilibrium point in sequence [5][6][7]. Actually, a number of muscles are concerned with each joint, when and which muscles are activated is largely dependent on the situation. According to the minimum torque change model, that optimal motor command (torque and forces) are directly obtained from the inverse dynamics of the musculo-skeletal system [8][9].

Suppose that the subject holds an object. When the subject lifts the object by himself, the hand supporting the weight remains nearly stationary after lift-off. However, when another person lifts the object suddenly, the hand moves upward. The torque supporting the weight is not required any more once the object is lifted. In the latter case, however, the subject does not know the lift-off timing; therefore, the CNS cannot adjust the torque control because the propagation error in the sensorimotor system and visual feedback exist, because of this error the hand moves upward. In the former case, because the subject knows the lift-off timing he can adjust the torque accordingly. As a result, the subjects' hand remains stable. In this case, the CNS has to send the motor command (muscle activation) to the agonist and antagonist muscles before lift-off, since it takes at least 100ms for muscles to generate tension force after receiving the commands from the CNS.

We hypothesized that muscle activation level under the self-lifting task would increase just before the lift-off. This would suggest that the CNS simultaneously controls not only the equilibrium point and torque, but also the impedance properties of the skeletal muscle as a motor command in postural control. From this result, because the stiffness level alters during posture control, the EP would be also controlled complicatedly.

This paper aims to investigate the role of the preparatory muscular activation of the musculo-skeletal system against the alteration of the environment. To quantify the changes in human arm viscoelasticity on the vertical plane before and just before the lift-off, we examined the relationship between the perturbation and a time-varying muscle co-activation level by utilizing the surface electromyographic studies.

In section 2 of this paper, we describe the formulation of a musculo-skeletal system. Section 3 outlines the experimental procedures used to collect and process of the experimental data. Section 4 shows experimental results in joint torque estimation, time-varying joint torque and stiffness during the posture control. In section 5, we discuss the impedance controllability of the musculo-skeletal system as the motor command from the experimental results.

II. The formulation of the musculo-skeletal system

The CNS drives and controls the skeletal muscle, which in turn is responsible for posture maintenance and voluntary movement. The CNS sends a motor command to each muscle via α motor neurons. These signals activate the muscle contraction(muscle tensions), which results in joint torque.

Most joints in the human arm are crossed by a number of

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muscles. With regard to a joint, there are extensor muscle group, which operate the direction of lengthening and flexor muscle group, which operate the direction of shortening. When the extensor and flexor muscles are relaxed simultaneously, the joint stiffness decreases, and the joint becomes easily influenced by external forces. On the other hand, when the extensor and flexor muscles are activated simultaneously, joint stiffness increases. This means that it is possible to adjust the impedance properties around the joint by changing the muscle activation level while maintaining the same torque. As it were, we can suggest that the impedance properties of the musculoskeletal system would contribute to adjustment of posture control.

In order to quantify the changes in human arm viscoelasticity, we measured surface EMG signals. We treat EMG signals as a record of the motor commands to the muscles, since we cannot directly measure the motor neuron activity. The EMG activity is a reasonable reflection of a firing rate of a motor neuron. Actual EMG activity was transformed by a linear, second order, low pass filter. The transformed signal is called 'quasi-tension', because it seems to correlate highly with the true muscle tension. We also used mathematical approaches to estimate the joint stiffness from the quasi-tension levels[11][12].

1. The relationship between motor command and muscle tension

A nonlinear relationship exists between muscle-exerted tension and motor commands. Muscle tension is related to muscle activity(firing rate) through a sigmoid function. This non-linearity is caused not only by the firing rate-tension relationship but also recruitment of α motor neurons. Moreover, there are two nonlinear relationships: one between muscle tension and muscle length, the other between muscle tension and muscle contractile velocity[10]. The first is called the length-tension curve, which describes how muscle tension increases with length even if the motor command does not change. The slope of this curve represents the elastic coefficient of the muscle. The second is called velocity-tension curve and describes how muscle tension decreases with contractile velocity for a constant motor command. This curve represents the characteristics of viscosity. Based on these characteristics, muscle can be modeled similar to a spring. Variation in motor command signal affects the elastic and damping coefficients and natural length resulting in a change in motor command, and can be modeled as follows:

$$\mathrm{T} = k(u,l)\Delta l(u,l) - b(u,l,\dot{l})\dot{l} \tag{1}$$

Where $u, l, \dot{l}$ denote motor command, muscle length, and muscle contractile velocity respectively. $k(u,l)$ $b(u,l,\dot{l})$ and $\Delta l(u,\ l)$ denote elastic and damping coefficients, and the stretch length from the natural length. In our studies, experiments are performed in static conditions, so we can ignore the damping term. As a result, muscle tension T can be determined from muscle stiffness k and the stretch length of each muscle as follows.

$$\mathrm{T} = k(u,l)\Delta l(u,l) \tag{2}$$

2. The relationship between EMG signals and quasi-tensions

Surface EMG response are spatiotemporally convoluted action potentials of the muscle membranes and involve not only descending central motor commands but also reflex motor commands generated from sensory feedback signals. There have been considerable efforts to estimate muscle force from surface EMG signals. From these previous studies in medical electronics and biological engineering, it can be expected that low-pass filtered EMG signals reflect the firing rate of α motor neurons as high frequency components of EMG reflect the shape of individual action potentials, while low-frequency components reflect the firing frequency of motor nerve fibers. In neurophysiological studies, it was found that a second-order, low-pass filter was sufficient for estimating muscles forces from the nerve impulse train. The relationship between the EMG input signal and T(quasi-tension) the output signal can be represented as an FIR filter.

$$\hat{\mathrm{T}}(t) = \sum_{j=1}^{n} h_j \cdot EMG\ (t - j + 1) \tag{3}$$

Where h_j and EMG represent the filter and EMG signals. $\hat{\mathrm{T}}(t)$ denotes 'quasi-tensions', and j is the number of discrete time. EMG is actually the digitally rectified, integrated, and filtered signal. The second-order frequency response of the filter H(s) is represented as follows:

$$H(s) = \frac{w_n^{\,2}}{(s^2 + zw_n s + w_n^{\,2})} \tag{4}$$

Where w_n and z denote natural frequency and damping coefficient, respectively. The impulse response of the function in (4) is

$$h(t) = a \times (\exp^{-bt} - \exp^{-ct}) \tag{5}$$

The coefficients h_j in Eq. (3) can be acquired by digitizing $h(t)$ with the given coefficients a, b, and c.

3. The relationship between quasi-tensions and joint torques

The joint torque is generated by the torque difference between the extensor and the flexor muscles and is dependant on the muscle tensions and moment arms. The joint torque τ is defined as follows

$$t_i = \sum_j A_{ij}(q) T_j \tag{6}$$

$$A_{ij}(q) \le 0 \ \text{; For extensor,} \quad A_{ij}(q) \ge 0 \ \text{; for flexor}$$

Where i and j denote the joints and muscles, and $A_{ij}(q)$ and T denote the moment arm and muscle tension. As mentioned above, we regarded quasi-tension as muscle tension.

We have developed a model to estimate the contribution of individual muscles to the total torque in a joint based on quasi-tensions. Six muscle groups are modeled. Each joint torque was estimated from the quasi-tensions and moment arms. In general, the moment arm varies nonlinearly with joint angle. In this study, because the experiments are executed in static condition, the muscle moment arm for three extensor muscles and three flexor muscles are assumed to be constant.

4. The relationship between the joint torque and the joint stiffness

Joint stiffness R is expressed as a differential operator that relates the small variation of joint torque to the small angular displacement q. The relationship between joint stiffness R and muscle stiffness S can be defined as follows[13].

$$R = -\frac{\partial t}{\partial q} = A \cdot S \cdot A^T \tag{7}$$

$$R = \begin{pmatrix} R_{ss} & R_{cj} \\ R_{cj} & R_{ee} \end{pmatrix} \tag{8}$$

$$A = \begin{pmatrix} d_1 & -d_2 & 0 & 0 & d_5 & -d_6 \\ 0 & 0 & d_3 & -d_4 & d_7 & -d_8 \end{pmatrix} \tag{9}$$

$$\begin{aligned} S &= diag\,[s_1, s_2, s_3, s_4, s_5, s_6] \\ &= diag\,[p_1 u_1 + s_{01} + \cdots p_6 u_6 + s_{06}] \end{aligned} \tag{10}$$

Here, d_i denotes the moment arm. We assume that muscle stiffness s_i is proportional to individual muscle activation u_i. s_1 and s_2 denote stiffness of shoulder mono-articular flexor and extensor, s_3 and s_4 denote stiffness of elbow mono-articular flexor and extensor, and s_5 and s_6 denote stiffness of bi-articular flexor and extensor, respectively. p_i denote positive constant coefficients, and s_i denotes the inherent stiffness value when the muscle is inactive. Off-diagonal terms of the joint stiffness matrix, R_{cj} are assumed to be equal, although the reflex components possibly may produce asymmetry in a stiffness matrix. Substituting Eq. (8)-(10) into Eq. (7), each term of joint stiffness R can be expressed as follows:

$$\begin{aligned} R_{ss} &= a_1 u_1 + a_2 u_2 + a_5 u_5 + a_6 u_6 + b_1 \\ R_{cj} &= a_9 u_5 + a_{10} u_6 + b_2 \\ R_{ee} &= a_3 u_3 + a_4 u_4 + a_7 u_5 + a_8 u_6 + b_3 \end{aligned} \tag{11}$$

$a_i u_i$ denotes effective muscle stiffness of the i-th muscle, that is, the contribution of j-th muscle activation to the corresponding joint stiffness. In our study, a_i and b_i are constant because the moment arm is assumed to be constant.

III. Experimental procedures

One healthy subject, 28 years old, participated in this study. The subject were seated on a chair and asked to position his shoulder, elbow and wrist on a vertical plane.

Shoulder and elbow muscle EMGs and limb kinematics were captured with an OPTOTRAK position sensing system. Four markers were placed on the hand, elbow, shoulder and the object and sampled at 200Hz. The marker position data were filtered digitally with a zero-lag Butterworth low-pass filter with a cutoff frequency of 20Hz. The marker data were

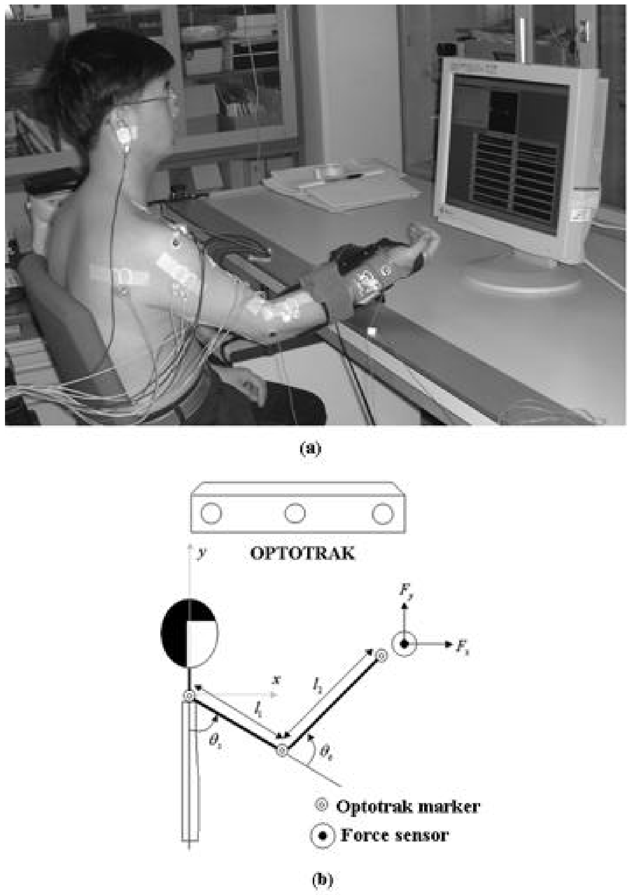

Fig. 1. Experimental Setup.

used to detect the hand position and the lift-off timing. Figure 1 shows the experimental settings.

The hand force was measured by a force-torque sensor and filtered at an upper cut-off frequency of 20Hz by the hardware. We measured surface EMGs from six muscles representative of mono-and bi-articular flexor and extensor muscles spanning the elbow and shoulder joints shown in fig. 2. Two shoulder mono-articular muscles, activity in the pectoralis mi-

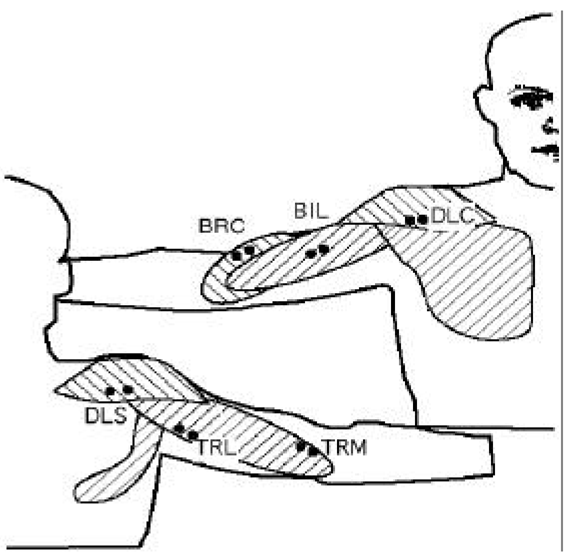

Fig. 2. Electrode positions in EMG measurements.

nor(flexor), and the infraspinatus(extensor) were measured. For elbow mono-articular muscles, activity in brachioradialis (flexor) and the lateral head of triceps of brachii(extensor) were measured. For bi-articular muscles, activity in the biceps brachii(flexor) and long head of triceps brachii (extensor) were measured. The EMG signals were recorded by using pairs of silver-silver chloride surface electrodes in bipolar configuration. Each signal was filtered and sampled at 2000Hz with 12bit resolution. Each electrode had a 10-mm diameter and was separated by approximately 15mm. The signal was digitally rectified, integrated for 0.5ms, sampled at 200Hz, and finally filtered. The experiments were performed in the following order.

1) isometric force generation
2) EMG signal measurements during relaxed condition in the same posture with 1)
3) co-contraction level measurement without hand force
4) lifting task with several weighed objects

1. Experiment 1: isometric force generation

In the first experiment, to calculate joint torques from the limb kinematics, the subject held the tip of a force-torque sensor and executed force in an upper direction. Four tests were performed and each time the monitor showed a range of 0 to 15[N] on the vertical plane. The EMG levels were also measured from six muscle groups during this condition. The shoulder and elbow joint angle data were calculated from the OPTOTRAK position data. Shoulder and elbow joint torques were calculated from the measured hand force within the vertical plane, multiplied by the transpose of the jacobian of the coordinate transformation.

2. Experiment 2: lifting task

These measurements of arm positions and EMG signals were simultaneously continued during the lifting task using the same method as in experiment 1. The experiments consist of two types of perturbations; one is lifting an object by another person, the other lifting the object by himself. We asked the subject to hold the object while varying the object weights: 0.5Kg, 1.0Kg, and 1.5Kg. Each step was performed four times and accomplished in the following order.

1) Another person lifts the 500g-weighed object.
2) The subject lifts the same weight object by himself.
3) Another person lifts the 1000g-weighed object.
4) The subject lifts the same weight object by himself.
5) Another person lifts the 1500g-weighed object.
6) The subject lifts the same weight object by himself.

Fig. 3 shows the time-varying hand position data from the OPTOTRAK and the quasi-tension data of six muscles under the self-lifting task.

IV. Experimental results

1. Joint torque and moment arm estimation

Because a nonlinear increase in EMG signal with muscle force has been reported in high muscle activation, we separated the data according to the contraction level (Basmajian and De Luca 1985). To specify the relationship among EMG signals, quasi- tension, and joint torques under isometric conditions, the measurements in experiment 1 were first estimated from surface EMG signals using mathematical procedures.

We estimated the moment arm parameters using the isometric force regulation data and quasi-tension data from EMG signals. We first substituted the calculated joint torques and quasi-tension into Eq(6), and then calculated the moment arm parametersby utilizing the least-square error fitting. A constrained optimization method using MATLAB software was used to estimate the parameters that satisfied Eq. (6). Table 1 shows the estimated moment arm parameters. The correlation coefficient between calculated and estimated moment arm parameters was 0.9331. We then reconstructed the estimated joint torques by substituting these parameters and quasi-tension data from six muscles into Eq. (6). Figure 4 shows the calculated and the estimated joint torques.

Table 1. Estimated moment arm parameters

shoulder joint		elbow joint	
a1	-0.1037	a5	-0.2645
a2	0.0241	a6	0.0349
a3	-0.1528	a7	-0.4392
a4	0.0537	a8	0.0715

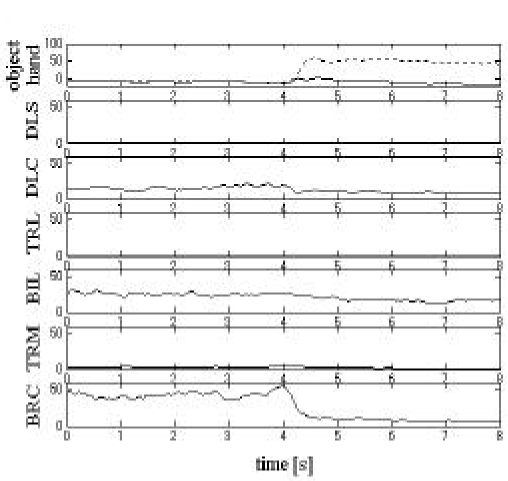

Fig. 3. Hand displacement and quasi-tensions.

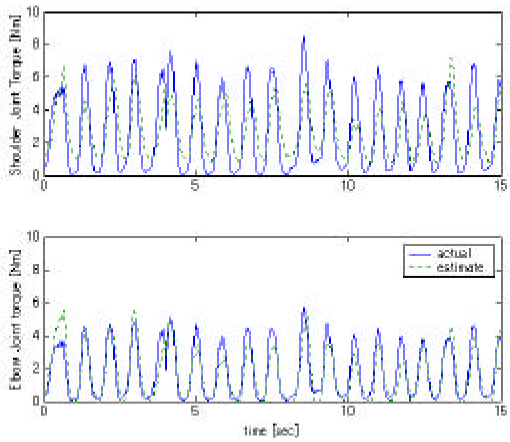

Fig. 4. The calculated and estimated joint torques.

2. Estimated joint torque and stiffness index

In our experiments, because moment arm parameters can be assumed to be constant, joint stiffness for shoulder and elbow can be expressed as a linear sum of quasi-tensions. To estimate joint stiffness index, we get an absolute value to the moment arm parameters in Table 1. Figure 5 shows the observed time-varying patterns of stiffness index and elbow joint torque when the subject lifts the object by himself. A solid line and a dotted line represent stiffness index and joint torque respectively. We used a curvature index to detect the lift-off timing from OPTOTRAK data. A horizontal axis denotes time interval, and 0 stands for the lift-off timing. Fig.5, from top to bottom, represents the experimental results for the 0.5Kg, 1Kg and 1.5Kg weighed objects respectively.

To examine the changes of the stiffness level and torque between just before the lift-off and states of holding object, we fixed –0.5~0[sec] for the section of just before lift-off (section b), and –3.0~-1.5[sec] for the section holding states (section a). Table 2shows the percentage ratios of the joint torque and stiffness index in section a and b by fixing 100% on the basis of sectiona.

As compared with incremental ratios of the joint torque, wecould observe that the stiffness level increased about 130% just before the lift-off. We call this characteristic of the musculo-skeletal system ‘pre-contraction’.

Table 2. Mean and standard deviation of the joint torque and stiffness index

object weight	joint torque		stiffness index	
	a	b	a	b
0.5Kg	100%	121%	100%	130%
1.0Kg	100%	109%	100%	126%
1.5Kg	100%	103%	100%	142%

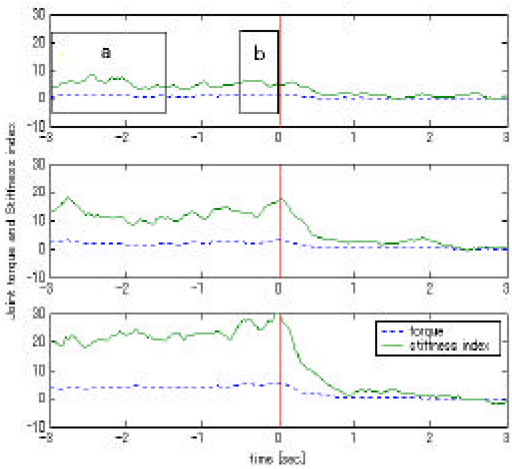

Fig. 5. Estimated joint torque and stiffness index.

V. Discussion

We have showed evidence that the impedance property of the musculo-skeletal system may contribute to the posture maintanance as a sort of motor command. We have examined the relatioship between EMG signals and joint stiffness against the alteration of the environments. Joint torque and stiffness index have been estimated from surface EMG signals using a mathematical procedure.

From the experimental results, we found, that when another person lifts the object suddenly, because the subject didn’t know the lift-off timing, as a result, the posture control is executed by decreasing the muscle tension and torque after the lift-off. On the other hand, when the subject lifts the object personally, we could observe that the subject increased the stiffness level just before the lift-off with non-variance in torque value. We call this characteristic of the musculo-skeletal system ‘pre-contraction’.

From this result, because the stiffness level alters during posture control, the EP is also controlled complicatedly. And the CNS as a sort of motor command simultaneously would control not only the EP and torque but also joint stiffness as well.

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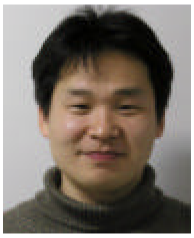

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