

# Pinpointed Muscle Force Control Using a Power-assisting Device: System Configuration and Experiment

Ming Ding, Jun Ueda and Tsukasa Ogasawara

**Abstract**—The demand for rehabilitation robots is increasing for the upcoming aging society. Power-assisting devices are considered promising for enhancing the mobility of elderly and disabled people. Other potential applications are for muscle rehabilitation and sports training. The main focus of this paper is to control the load of selected muscles by using a power-assisting device, thus enabling “pinpointed” motion support, rehabilitation, and training by explicitly specifying the target muscles. By taking into account the physical interaction between human muscle forces and actuator driving-forces during power-assisting, the feasibility of this muscle force control is analyzed as a constrained optimization problem. A prototype power-assisting device driven by pneumatic rubber actuators is developed. A control system is developed with a graphical user interface that provides an easy operation to designate desired forces for target muscles. The validity of the method is confirmed by experiments by measuring surface electromyographic (EMG) signals for target muscles.

## I. INTRODUCTION

For the upcoming aging society, power-assisting devices for enhancing the mobility of elderly and disabled people are considered important. The power-assisting devices are designed as wearable mechanisms with a number of actuators, which reduce the burden of the wearer’s muscle loads such as for walking, running, and carrying a heavy load. Various power-assisting devices have been developed [1], [2], [3], [4], [5], [6].

We have proposed the concept of **Pinpointed Muscle Force Control (PMFC)** [7] for a more sophisticated application of power-assisting devices. The basic concept is to use a power-assisting device not to merely assist joint torques but to assist the load of arbitrarily selected muscles. Power-assisting during motion causes an interaction between human muscle forces and actuator driving-forces where both of them have a number of degrees of freedom. This implies that an appropriate control of actuators, by taking this interaction into account, may enable more accurate assisting at the level of individual muscle forces. Consider an application to individual muscle rehabilitation. It is considered effective to selectively assist/resist weakened muscles in order to gain the functionality of them, and not to impair other healthy muscles. This approach may be applicable to sport training where only specific muscles are of interest for training by applying the load to them.

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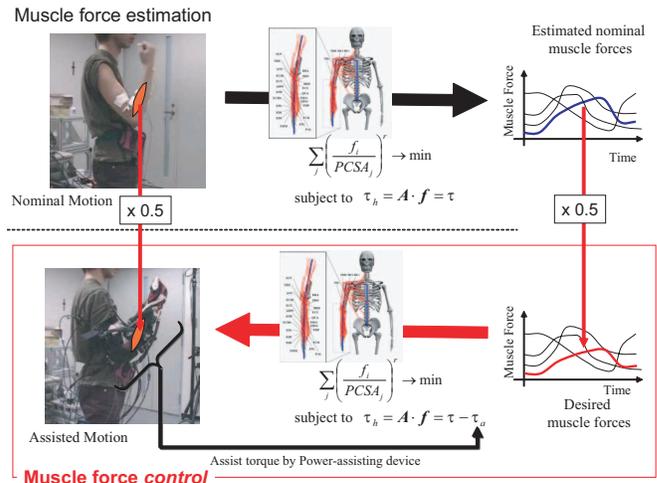


Fig. 1. Concept of Muscle Force Control

This muscle force control is formulated as a constrained optimization problem. The actuator driving-forces are calculated based on Crowninshield’s cost minimizing law [8] for estimating the distribution of human muscle forces. A mathematical analysis obtains a closed-form solution for a feasible set of desired forces for target muscles. This analysis is more rigorous than the numerical iterative method presented in [7].

A control system named **Muscle Assistant System (MAS)** has been developed that is composed of a 3-dimensional motion capturing system, power-assisting device, and calculation software. Further, the software includes a musculoskeletal human model, feasibility analyzer, power-assisting device controller, and graphical user interface. A prototype power-assisting system with pneumatic actuators has been developed for assisting human upper-right limb. Also, a musculoskeletal model with 51 muscles for the limb has been developed. The validity of Pinpointed Muscle Force Control is confirmed by experiments by measuring surface electromyographic (EMG) signals of target muscles.

## II. CONCEPT OF PINPOINTED MUSCLE FORCE CONTROL

Our basic concept for novel intelligent power-assisting is not to assist at the level of joint torques but to assist at the level of individual muscle forces. This “pinpointed” control of muscles enables us to modify the load of selected target muscles by applying torques from a wearable, active robotic device. Consider a case where a human subject wears an exoskeleton or a power-assisting device driven

by actuators. The exoskeleton robots are designed to apply forces or torques to the interface between the robot and the wearer. Note that the torques created by the actuators are transmitted to the wearer's body through the joints. The joint-level effect of assisting is relatively easy to examine, by analyzing the interaction between two rigid link mechanisms, i.e., the human body and assisting device.

The effect on specific muscles needs to be examined for effective power-assisting such as gait rehabilitation [9]. The analysis of the effect at the level of individual muscles from a given set of joint torques is fundamentally an ill-posed problem since a human body has a redundant number of muscles than the number of joints. This problem may be solved by utilizing a muscle force estimation method proposed in physiology [8] where the distribution of human muscle forces is assumed to be subjected to a physiologically-based criterion function. The upper part of Fig. 1 illustrates this scheme where the distribution of subject's muscle forces is estimated from posture measurement. As far as the human joint torques are obtained, this muscle force estimation works for both cases with and without power-assisting.

The concept of pinpointed muscle force control is to control a power-assisting device such that a desired set of muscle forces is realized as illustrated in the bottom of Fig. 1. This method aims at arbitrarily modifying the load of selected target muscles, thus enabling "pinpointed" support, rehabilitation, or training of target muscles. The bottom part of Fig. 1 shows the idea of the muscle force control; the flow is basically the inverse of muscle force estimation, where desired muscle forces are given, e.g., based on nominal muscle forces without assisting, then the joint torques that the power-assisting device generates are calculated. However, this muscle force control at the level of individual muscles is not straightforward. Excluding direct stimulation of individual muscles, the distribution of muscle forces are indirectly controlled through the modification of a limited number of joint torques by power-assisting mechanisms. In addition, the muscle force control is not totally arbitrary since human muscle forces are subject to a certain physiology-based rule such as Crowninshield's law, therefore the class of feasible muscle needs to be analyzed. The key idea for this pinpointed muscle force control is to formulate the problem as a constrained optimization problem considering the physical interaction between human muscle forces and actuator driving-forces.

### III. MUSCLE ASSISTANT SYSTEM

#### A. System Overview

A control system named **Muscle Assistant System (MAS)** has been developed for pinpointed muscle force control. The system configuration is shown in Fig. 2(a). The system composed of 4 modules: human posture measurement module, muscle force estimation module, muscle force control module, and power-assisting device control module. A graphical user interface provides an easy operation to designate desired forces for target muscles, and to view the resultant distribution of the muscle forces.

1) *Posture Measurement Module*: A target motion is measured by a motion capture system (Mac3D system, Motion Analysis Corporation). Measurement software (EvaRT) reproduces the posture of the subject from 3 dimensional positions of the markers attached on the subject and calculates joint angles, velocities, and accelerations.

2) *Muscle Force Estimation Module*: Joint torques are calculated by substituting the obtained joint data for a musculoskeletal human model that will be described below. This skeletal model provides the moment arms and lengths of muscles, then a physiologically based criterion of muscle forces [8] is applied to estimate the human muscle forces for a given posture. This estimation is solved as a quadratic programming problem using MATLAB. The obtained muscle forces, hereafter called nominal muscle forces, are used as the basis for the designation of muscle forces in the following module.

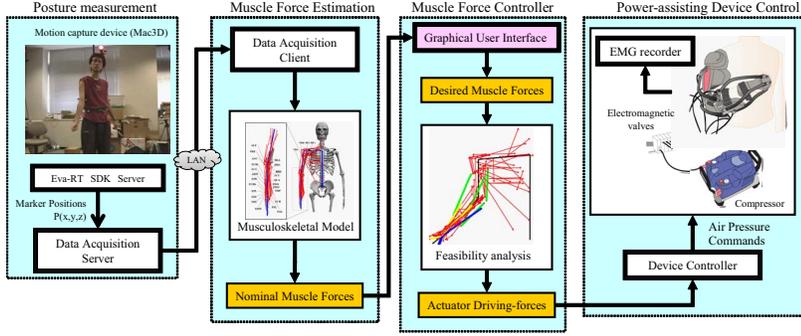
3) *Muscle Force Control Module*: An operator designates desired muscle forces for target muscles using a graphical user interface as shown in Fig. 2(b). The rates of change for target muscles are designated based on the nominal muscle forces by a slide bar interface. An important feature of this module is that the feasibility of the designated muscle forces is checked before execution of actual power-assisting based on a physiological criterion for human muscle force generation. The detail of this analysis will be described later. If the given set of desired muscle forces is feasible, the resultant human muscle forces are calculated for all the muscles including non-target muscles.

4) *Device Control Module*: The human joint torques when assisted are calculated from the resultant muscle forces. By subtracting the resultant human joint torques from the nominal joint torques, the torques that need to be generated by the power-assisting device are calculated. This device control module calculates the pressure for each pneumatic actuator to realize the resultant torques. The pressure reference commands are then sent to electropneumatic regulators and execute the assist.

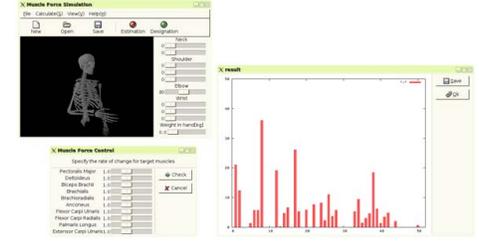
#### B. Power-assisting Device driven by Pneumatic Actuators

A power-assisting device using pneumatic actuators shown in Fig. 3(a) has been developed for modifying or *controlling* muscle forces of human upper-right limb [7]. Although pneumatic actuators are known to have slow responses, the compactness and lightwightness excluding compressors are considered suitable for developing a device with multiple degrees of freedom (DOF). This device controls 4 DOF motion of the right arm by 8 actuators, including 1 DOF of the elbow and 3 DOFs of the wrist. Both ends of each actuator are attached to plastic frames which are then attached to the body by Velcro tapes. Unlike other exoskeleton mechanisms, this device does not have rigid frames for safety reason. This device may be modeled as shown in Fig. 3(b) that includes the information on the lengths and moment arms for individual actuators.

A pneumatic actuator shown in Fig. 3(a) with 20 [mm] diameter, maximum pressure of 0.4 [MPa], and maximum



(a) System configuration



(b) Graphical user interface

Fig. 2. Muscle Assistant System



(a) Power-assisting device

(b) Device model

Fig. 3. Power-assisting Device and Device Model

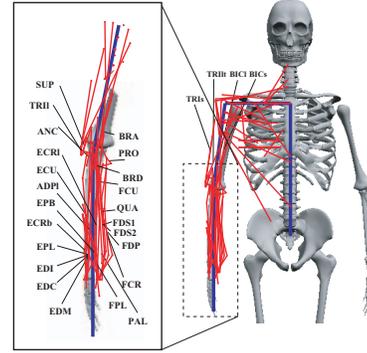


Fig. 4. Musculoskeletal Model of Upper Right Limb

force of 60 [N] is used. This actuator contracts when pressurized by a compressor controlled by an electropneumatic regulator. For simplicity the characteristic of the pneumatic actuator is modeled by linear approximation. The force  $p$  created by the actuator is calculated simply by Hooke's law:

$$p = \eta(l - l_{\text{free}}(P)), \quad (1)$$

where  $P$  is the air pressure controlled by an electropneumatic regulator,  $l_{\text{free}}(P)$  is a neutral length of the actuator as a function of the air pressure, and  $l$  is the actual length of the actuator.  $\eta$  is a spring constant of the actuator.  $\eta = 6250[\text{N/m}]$  is applied as the stiffness of the actuator. By using (1), the air pressure  $P$  is calculated for a desired force  $p$  and the current length of the actuator  $l$  that is calculated by posture measurement of the device.

### C. Musculoskeletal Human Model and Muscle Force Estimation

A musculoskeletal model of human upper right limb has been developed [7] as shown in Fig. 4. This model consists of 5 rigid links with 13 joints corresponding to the waist, neck, shoulder, elbow, and wrist. 51 muscles of the upper-right limb presented in Table I are modeled by massless wires [10] [11]. Points of muscle attachment (origins and insertions) are determined from anatomical data [10]. The validity of the musculoskeletal model has been evaluated in terms of muscle moment arms [7].

Suppose that the human musculoskeletal model has  $M$  joints and  $N$  muscles. Let  $\mathbf{f} = [f_1, \dots, f_N]^T \in \mathbb{R}^N$  be a contraction force vector of the human muscles. The relation between the human joint torque  $\boldsymbol{\tau}_h \in \mathbb{R}^M$  and  $\mathbf{f}$  is given by

$$\boldsymbol{\tau}_h = \mathbf{A}\mathbf{f} = \begin{bmatrix} a_{11} & \cdots & a_{1N} \\ \vdots & \ddots & \vdots \\ a_{M1} & \cdots & a_{MN} \end{bmatrix} \begin{bmatrix} f_1 \\ \vdots \\ f_N \end{bmatrix} \quad (2)$$

where  $\mathbf{A}$  is a moment arm matrix of the muscles. The element  $a_{ij}$  denotes the moment arm of muscle  $j$  for joint  $i$ .  $a_{ij} = 0$  is given if  $f_j$  does not affect on joint  $i$ . Note that  $\mathbf{A}$  is a function of joint angles and provided by the musculoskeletal model in Fig. 4.

Generally human body has a redundant number of muscles than the number of joints, i.e.,  $N \gg M$ , which makes the estimation of muscle forces  $\mathbf{f}$  from joint torques  $\boldsymbol{\tau}_h$  an ill-posed problem. Crowninshield's cost function [8] is to solve this problem by minimizing a physiologically based criterion  $u(\mathbf{f})$  as follows:

$$u(\mathbf{f}) = \sum_{j=1}^n \left( \frac{f_j}{\text{PCSA}_j} \right)^r \rightarrow \min \quad (3)$$

$$\text{subject to } \begin{cases} \boldsymbol{\tau}_h = \mathbf{A}\mathbf{f} \\ f_{\min_j} \leq f_j \leq f_{\max_j} (j = 1, \dots, N) \end{cases},$$

where  $\text{PCSA}_j$  is the physiological cross sectional area (PCSA), and  $f_{\max_j} = \varepsilon \cdot \text{PCSA}_j$  is the maximum muscle

TABLE I  
LIST OF MUSCLES IN MUSCULOSKELETAL MODEL

No.	Muscle Name	No.	Muscle Name	No.	Muscle Name	No.	Muscle Name
1	Levator Scapulae	14	Latissimus dorsi medial	27	Brachialis	40	Pronator Quadratus
2	Pectoralis major up	15	Latissimus dorsi lower	28	Brachioradialis	41	Abductor Pollicis Longus
3	Pectoralis major low	16	Subscapularis	29	Triceps long	42	Pronator teres
4	Pectoralis minor	17	Deltoideus anterior	30	Triceps lateral	43	Extensor Carpi Ulnaris
5	Subclavius	18	Deltoideus lateral	31	Triceps short	44	Extensor Carpi Radialis
6	Serratus ant.upper	19	Deltoideus post	32	Anconeus	45	Extensor Carpi Radialis
7	Serratus ant.lower	20	Supraspinatus	33	Flexor Carpi Ulnaris	46	Common Digital Extensor
8	Trapezius upper	21	Infraspinatus	34	Flexor Carpi Radialis	47	Extensor Digiti Minimi
9	Trapezius medial	22	Teres major	35	Palmaris Longus	48	Extensor Indicis
10	Trapezius lower	23	Teres minor	36	Flexor Digitorum Superficialis	49	Extensor Pollicis Longus
11	Rhomboids upper	24	Coracobrachial	37	Flexor Digitorum Superficialis	50	Extensor Pollicis Brevis
12	Rhomboids lower	25	Biceps long	38	Flexor Digitorum Profundus	51	Supinator
13	Latissimus dorsi upper	26	Biceps short	39	Flexor Pollicis Longus		

force for muscle  $j$ .  $\varepsilon = 0.7 \times 10^6 [\text{N/m}^2]$  was given by Karlsson [12] and PCSA $_j$ 's are found in [13].  $f_{\min j} = 0, \forall j$  is used. Also, a quadratic cost function, i.e.,  $r = 2$ , is used for simplicity. See [8] for the choice of  $r$ .

#### IV. PINPOINTED MUSCLE FORCE CONTROL: OVERVIEW

Consider a motion assisted by a power-assisting device. Let  $\tau \in \mathbb{R}^M$  be a total joint torque vector needed for a target motion. At joint level, the resultant joint torque  $\tau$  is simply the sum of human torques and assist torques represented by

$$\tau = \tau_h + \tau_a \quad (4)$$

where  $\tau_a \in \mathbb{R}^M$  is a torque generated a power-assisting device. The joint torque  $\tau_a$  of the assisting device is given by

$$\tau_a = \mathbf{E}\mathbf{p} = \begin{bmatrix} e_{11} & \cdots & e_{1L} \\ \vdots & \ddots & \vdots \\ e_{M1} & \cdots & e_{ML} \end{bmatrix} \begin{bmatrix} p_1 \\ \vdots \\ p_L \end{bmatrix} \quad (5)$$

where  $L$  is the number of pneumatic actuators,  $\mathbf{p} = [p_1, \dots, p_L]^T$  is the driving force vector of the actuators, and  $\mathbf{E}$  is the moment arm matrix of the actuators provided by Fig. 3(b).

Assume that a wearer is able to perform the same target motion during power-assisting, i.e., the time series data for  $\tau$  and  $\mathbf{A}$  with assisting are the same as for the nominal motion. Therefore, the power-assisting device modifies the relative ratio between  $\tau_h$  and  $\tau_a$  for  $\tau$  unchanged. According to (4) and (5), the equality condition of (3) is changed by  $\mathbf{p}$ , resulting in the indirect control of  $\mathbf{f}$  by  $\mathbf{p}$ .

The following four-step algorithm as shown in Fig. 5 is proposed to realize this muscle force control:

**[Step 1] Posture measurement of a target motion:** First, a target motion is measured by a motion capture system.

**[Step 2] Muscle force estimation for unassisted motion:** From the measured joint torques and arm posture, the nominal muscle force  $\mathbf{f}_0$  for the target motion without using the power-assisting device is calculated by Crowninshield's method.

**[Step 3] Designation of desired muscle forces and feasibility analysis:** Based on  $\mathbf{f}_0$  in Step 2, the desired forces

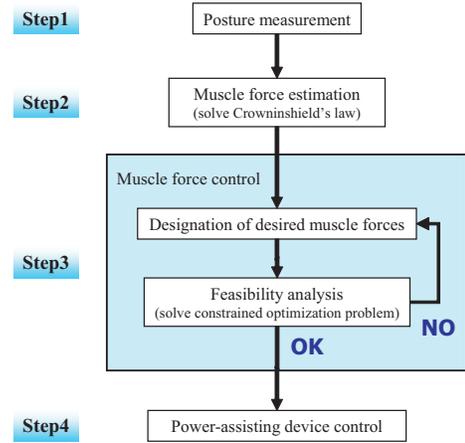


Fig. 5. Muscle Force Control Algorithm

for target muscles  $\mathbf{f}_{d_o} = [f_{d_o,1}, \dots, f_{d_o,N_o}]^T \in \mathbb{R}^{N_o}$  are designated as

$$f_{d_o,j} = \gamma_j f_{0_o,j} \quad (j = 1, \dots, N_o) \quad (6)$$

where  $N_o$  is the number of the target muscles, and  $\gamma_j$  is the rate of change for muscle  $j$ . The feasibility is checked in consideration of a constrained optimization problem according to Crowninshield's method. The effect on the other muscle forces are then calculated if  $\mathbf{f}_{d_o}$  is feasible, and the resultant device torque  $\tau_a$  that realizes  $\mathbf{f}_{d_o}$  is calculated.

**[Step 4] Control of power-assisting device:** The driving-forces and pressure commands for the pneumatic actuators are calculated to generate  $\tau_a$ .

As described earlier, the power-assisting device does not directly control human muscle forces but merely modifies human joint torques, which is equivalent to the modification of the equality condition for (3). Recall that the resultant distribution of human muscle forces is determined according to Crowninshield's cost function. In other words, the proposed pinpointed muscle force control is an indirect control of muscle forces by an appropriate modification of the equality condition for cost function optimization. This problem can be solved as a constrained optimization problem. MATLAB Optimization Toolbox may be useful for numerical solution.

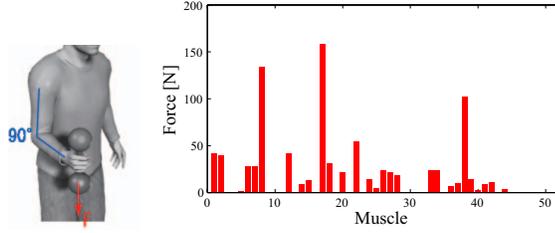


Fig. 6. Static Posture for simulation and nominal muscle forces.

An analysis applying Kuhn-Tucker theorem [14] provides a mathematically rigorous solution.

## V. SIMULATION

First, the validation of the muscle force control is confirmed by simulation. A posture shown in Fig. 6 is considered as an example where the elbow joint angle is  $90^\circ$  and an external downward force (19.8[N]) is applied to the hand by holding an iron dumbbell. Figure 6 shows the normal force  $f_0$  for this posture, where the horizontal axis of the graph represents the number of muscles in Table I. The following two cases are examined: (A) control of a single muscle force and (B) simultaneous control of multiple muscle forces.

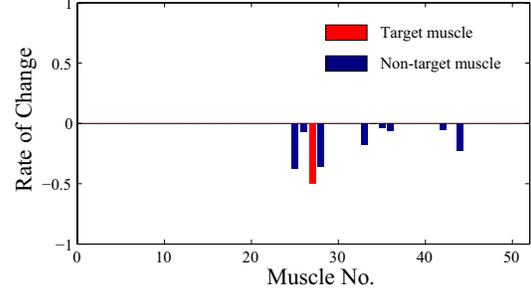
### A. Pinpointed Control of Single Muscle Force

In this example, Brachialis (No.27) is considered as a target muscle. The desired muscle force for Brachialis is given as a half of the normal muscle force, i.e.,  $f_{d_{27}} = 0.5 \times f_{0_{27}}$ . Using the remaining control DOFs, other muscle forces are controlled to minimize the variation of change. Figure 7(a) shows the incremental rate of change of the controlled muscle forces from the nominal muscle forces. For example, if a nominal muscle force and the controlled muscle force are the same, this graph shows 0. Similarly, this graph shows  $-0.5$  if a controlled muscle force is a half of the nominal muscle force;  $0.5 - 1 = -0.5$ . Red bars represent the target muscles and blue bars represent the non-target muscles.

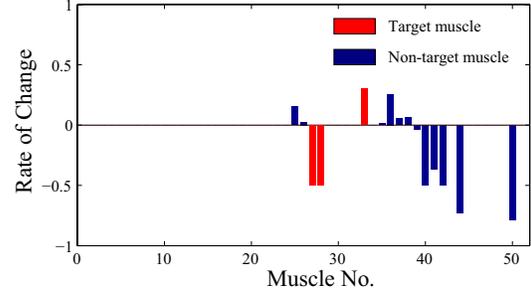
The anatomical function of Brachialis is flexion/extension of the elbow. This implies that a certain group of muscles having the similar function for the elbow may change in consequence of the control of Brachialis. As shown in the figure, Brachialis changed accordingly, showing  $-0.5$ . However, other 8 non-target muscles also changed due to the coupling among muscles although this effect has been minimally reduced.

### B. Simultaneous Control of Multiple Muscles

Figure 7(b) shows the result when multiple muscle forces are simultaneously controlled. In this example, desired force of Brachialis(No.27), Brachioradialis(No.28) and Flexor Carpi Ulnaris(No.33) are given as  $0.5 \times f_{0_{27}}$ ,  $0.5 \times f_{0_{28}}$  and  $1.3 \times f_{0_{33}}$ , respectively. Brachialis and Brachioradialis are for moving the elbow, and Extensor Carpi Ulnaris is for moving the wrist. This simultaneous control is challenging; Brachialis and Brachioradialis are assisted but Flexor Carpi



(a) Rate of change for single muscle force control: Brachialis (No.27) $\times 0.5$ .



(b) Rate of change for multiple muscle force control: Brachialis (No.27) $\times 0.5$ , Brachioradialis (No.28) $\times 0.5$ , Flexor Carpi Ulnaris (No.33) $\times 1.3$ .

Fig. 7. Simulation Results of Muscle Force Control

Ulnaris are resisted by increasing the load. The function of Brachialis and Brachioradialis is very similar, thus these two muscles are strongly coupled. Therefore, the same desired rate of change is given to them. The result shows that all the three target muscles have been adequately controlled. Similarly, several non-target muscles change due to the coupling among muscles.

## VI. EXPERIMENT

Muscle force control experiments for a posture shown in Fig. 8 are conducted by using the power-assisting device. The surface electromyographic signals (EMGs) are measured for validating the actual changes of muscle forces. The EMGs are used to check the tendency of change of muscle forces, i.e., increase or decrease. Note that EMGs are not accurate enough to measure the amount of change since the relation between the magnitude of muscle force and the one of the corresponding EMG is not necessarily linear.

The target muscles are Brachialis (No.27), Brachioradialis (No.28), and Extensor Carpi Ulnaris (No.43). The desired rates of change are given as shown in Table II. For example, Experiment A is to support only Extensor Carpi Ulnaris, Experiment C is to support only Brachialis and Brachioradialis, and Experiment E is the mixture of assisting and resisting. Since Brachialis and Brachioradialis are physiologically coupled, these two muscles are treated

TABLE II  
DESIRED RATES OF CHANGE FOR EXPERIMENTS

No.	Name	Experiment				
		A	B	C	D	E
27	Brachialis	x 1.0	x 1.0	x 0.5	x 0.5	x 0.5
28	Brachioradialis	x 1.0	x 1.0	x 0.5	x 0.5	x 0.5
43	Extensor Carpi Ulnaris	x 0.5	x 1.3	x 1.0	x 0.5	x 1.3

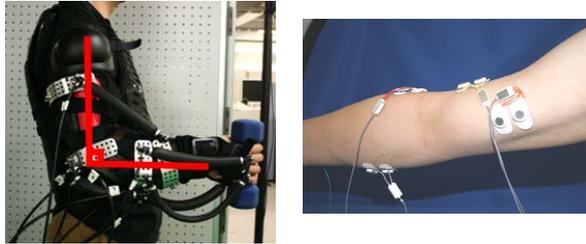


Fig. 8. Muscle Force Control Experiment: Posture and EMG measurement

as a group, giving the same rate of change. The feasibility for all the five experiments has been confirmed.

Each experiment is conducted for 6 male subjects. Figure 9 shows the results. The light gray bars show the desired changes of the target muscles and the dark gray bars show the changes of the measured EMGs based on the nominal cases without power-assisting. As shown in the graphs, all the tendencies of the change among the EMGs are as expected. For example, in Experiment D, all the target muscle forces reduced accordingly. Similarly, both Brachialis and Brachioradialis reduced, and Extensor Carpi Ulnaris increased in Experiment E, implying that even the mixture of assisting and resisting has been realized not only by simulation but also by experiment.

## VII. CONCLUSION

In this paper, a novel muscle force control algorithm has been proposed to modify the load of selected muscles by using a power-assisting device, thus enable the “pinpointed” motion support, rehabilitation and sport training. A prototype power-assisting system with pneumatic actuators has been developed for assisting human upper-right limb. The validity of the method has been confirmed by measuring surface EMG signals for static postures during power-assist. Future work includes improvement of the hardware, extension to the assist of dynamic motion, and more detailed clinical testing.

## REFERENCES

- [1] H. Kazerooni. Extender: A case study for human-robot, interaction via transfer of power and information signals. *Proc. IEEE Int. Workshop on Robot and Human Communication*, pages 10–20, 1993.
- [2] S. Lee and Y. Sankai. Power assist control for walking aid with hal-3 based on emg and impedance adjustment around knee joint. *Proc. IEEE/RSJ Int. Conf. Intelligent Robots and Systems*, pages 1499–1504, 2002.
- [3] M.A. Alexander, M.R. Nelson, and A. Shah. Orthotics, adapted seating and assistive devices. *Pediatric Rehabilitation*, pages 186–187, 1992.
- [4] H.I. Krebs, J. Celestino, D. Williams, M. Ferraro, B. Volp, and N. Hogan. A wrist extension for mit-manus. *Advances in Rehabilitation Robotics*, pages 377–390, 2004.
- [5] A. Toth, G. Arz, G. Fazekas, D. Bratanov, and N. Zlatov. Post stroke shoulder-elbow physiotherapy with industrial robots. *Advances in Rehabilitation Robotics*, pages 391–411, 2004.

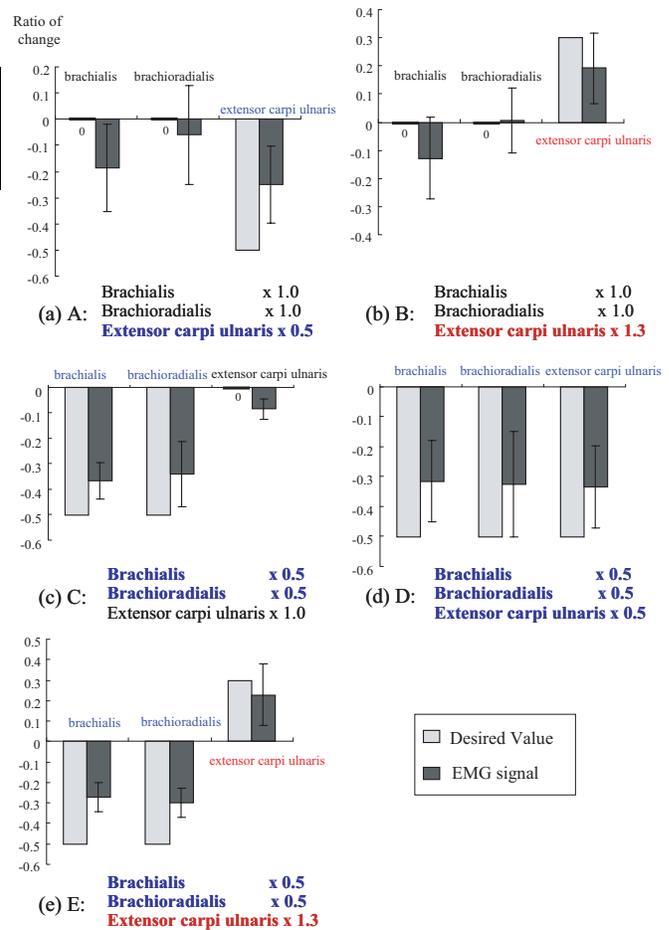


Fig. 9. Experimental Results: Rate of change

- [6] N. Benjuya and S.B. Kenney. Hybrid arm orthosis. *Journal of Prosthetics and Orthotics*, 2(2):155–163, 1990.
- [7] J. Ueda, M. Matsugashita, and T. Ogasawara R. Oya and. Control of muscle force during exercise using a musculoskeletal-exoskeletal integrated human model. In *10th International Symposium on Experimental Robotics (ISER2006)*, Rio de Janeiro, Brazil, July 6 to July 10, 2006.
- [8] RD. Crowninshield et al. A physiologically based criterion of muscle force prediction in locomotion. *J. Biomechanics*, 14:793–801, 1981.
- [9] Jan F. Veneman, Rik Kruidhof, Edsko E. G. Hekman, Ralf Ekkelenkamp, Edwin H. F., Van Asseldonk, and Herman van der Kooij. Design and evaluation of the lopes exoskeleton robot for interactive gait rehabilitation. *Ieee Transactions on Neural Systems and Rehabilitation Engineering*, 15(3):379–386, 2007.
- [10] W. Maurel et al. A case study on human upper limb modelling for dynamic simulation. *Computer Methods in Biomechanics and Biomechanical Engineering Gordon and Breach Science Publishers*, 2(1):65–82, 1999.
- [11] S. L. Delp et al. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Trans. Biomedical Engineering*, 37(8):757–767, 1990.
- [12] D. Karlsson et al. Towards a model for force predictions in the human shoulder. *J. Biomech*, 25:189–199, 1991.
- [13] MotCo project. <http://www.motoco.dir.bg/data/pcsa.html>.
- [14] D. Bersekas. *Nonlinear Programming*. 2nd edition, Athena Scientific, 1999.