MYOELECTRICALLY CONTROLLED HAND PROSTHESIS WITH NEUROMUSCULAR CONTROL SYSTEM DYNAMICS

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Abstract

The purpose of the present study is to develop a new type of myoelectric hand prosthesis which is provided with fundamental dynamic properties of the neuromuscular control system of finger muscles, in particular, mechanical properties of the muscle and of the myotatic reflex. These dynamics were estimated by performing stretch experiments with flexor pollicis longus muscles in man. It was showed that one of the characteristic features was an increase in mechanical impedance with an increase in activities of the muscle.

A one-degree-of-freedom prosthetic hand with those nonlinear dynamic properties were realized by utilizing position control system, force feedback and variable gain. The sum of and the difference between rectified, smoothed EMG signals of two antagonist muscles were used as control signals; an optimal smoothing filter was designed so that the smoothed signal is proportional to the isometric force of the muscle. Usefulness of the developed prosthesis was indicated; the prosthesis was smoothly controlled myoelectrically by the healthy subjects, and behaved softly like a natural limb, in response to disturbance force.

Key words;
myoelectric hand prothesis, servo controlled prosthesis, electromyogram, neuromuscular system dynamics, variable mechanical impedance

1. INTRODUCTION

Replacement of lost hand function is one of the most challenging problems of rehabilitation engineering. Since initial comment by Wiener regarding the use of bioelectric signals for control of devices, a variety of myoelectric prostheses have been so far developed and clinically utilized successfully 1-5). Most of the myoelectric prostheses moves in the on-off mode or the proportional mode according to the amplitude of EMG signal. Although much progress have been made, motor functions of myoelectric prostheses are still not to be compared with that of natural limb partly because they are designed with slight consideration to neural control mechanisms in a human limb. If the prosthesis possesses the same mechanism and mechanical properties as the neuromuscular control system, the amputee might be able to control the prosthesis more easily or to use subconscious control similar to that of the natural limb.

The purpose of the present paper is to develop such a new type of myoelectrically controlled prosthetic hand that is provided fundamental properties of the finger muscles in man and of the myotatic reflex. The prosthesis is expected to have two following advantages.
(1) The amputee could utilize almost the same subconscious control as he used before the amputation of hand. Consequently, brief training periods are required for the amputee to operate the prosthesis.
(2) Compliance (mechanical impedance) of the prosthesis as well vs velocity of the opening or closing is controlled voluntarily, vai activation patterns of EMG signals, by the amputee. The prosthesis thus developed is expected to fit for use in execting fine tasks or handling breakable materials.

This paper begins with representing a design principle of the prosthesis. Secondly, a myoelectric signal processor which is best for the control of the developed prosthesis is designed. Thirdly, dynamic properties of the neuromuscular control system in human finger muscles are estimated. Fourthly, servo controlled hand prosthesis with the estimated dynamics is implemented. Finally, myoelectric control experiments with healthy subjects are carried out to indicate usefulness of the developed prosthesis.

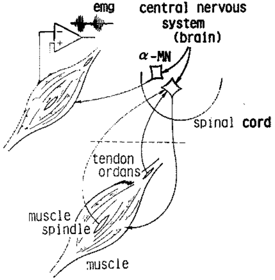

Fig. 1 Schematic drawing of neuromuscular control system. Functions of the muscle and of the myotatic reflex are replicated for the prosthesis, and command signals are picked up by EMG signals of remnants of the muscle.

2. DESIGN PRINCIPLE OF THE MYOELECTRIC HAND PROSTHESIS

2.1 Basic design scheme for the prosthesis

The human limb system includes a highly integrated network of multilevel controllers, actuators and receptors. Efferent impulses are sent from higher decision centers, via the spinal cord, to target muscle, followed by contraction of the muscle. Simultaneously, receptors continually provide the central nervous system with information necessary to moderate both voluntary and involuntary reflex responses. One of the most significant reflexes that play a major role in motor control is a myotatic reflex (stretch reflex), the arc of which consists of proprioceptors (muscle spindles and Golgi tendon organs), alpha motoneurons and muscles (see Fig. 1). It is well known that the myotatic reflex functions as being servo-assisted length control 6,7) and/or regulating stiffness of the motor servo (neuromuscular control system) 8).

As for the biological actuator, the muscle itself is flexible like a rubber, and produces required joint output torque by its contraction. Furthermore, viscoelasticity of the muscle varies with the contractile activities, so that the stiffness of joint is adjusted, via activation patterns of the muscle, by the higher centers so as to modulate the limb movement or to adopt against external disturbances.

These functions of the muscle and of the myotatic reflex are lost when the limb has been amputated (see broken line in Fig. 1). It is natural that these essential performance features are those which should be replicated by the powered prosthesis. The present study is along this line; that is, it attempts to develop a prosthetic hand which is provided partially with these fundamental functions and dynamic properties of the neuromuscular control system. In that case, command signals sent from the higher decision centers could be picked up by processing EMG signals of remnants of the muscles or their synergetic muscles.

2.2 Formulation of the neuromuscular control system

Dynamic behaviors of a pair **of antagonistic muscles** (flexor and extensor muscles) **are considered here. Each** constituent muscle is assumed identical **to ones for simplicity.** When the joint is maintained at **a certain angle, joint output** torque $P(t)$ is difference between **isometric torque of the** flexor $A_f(t)$ and that of the extensor $A_e(t)$. **If the joint** moves, torques evoked by the effects of **viscoelasticity of the** muscles as well as those evoked by **the** myotatic **reflex are** added. Denoting the transfer function of the **viscoelastic** properties by $G_1(s)$ and that of the myotatic reflex by $\exp(-Ls)G_2(s)$, a dynamic equation of the **neuromuscular control** system is expressed by

$$P(s)=A_f(s)-A_e(s)+G_x(s)\theta(s) \quad (1)$$

$$G_x(s)= G_1(s)+\exp(-Ls)G_2(s)$$

where θ is an angle of the joint and $G_x(s)$ **is the transfer** function representing dynamics of the neuromuscular control system and L is time delay of the myotatic **reflex arc. While** real dynamics of the system are **surprisingly** complex **with** nonlinearities and time delays 9), following simplest,useful description of $G_x(s)$ is utilized in the present **work as a first step.** The transfer function is derived by **investigating responses of the finger muscle to stretch; it will be indicated that the simple transfer function is sufficient enough to represent dynamic behaviours of the finger muscle as a first approximation (see Fig. 5).**

$$G_x(s)=K\ (\ 1 + T_1\ s)/\ (\ 1+ T_2\ s) \quad (2)$$

$$K= a(\ A_f+A_e)+ K_o \quad (3)$$

Note that the gain K is not constant, but proportional to the sum of the exerted isometric torques of the antagonistic muscles, A_f+A_e. It implies that mechanical impedance of the neuromuscular system varies with activities of the muscles. The dynamics at the resting state are also expressed by Eq.(2) with the gain K_o.

2.3 Control system of the prosthesis

The control system which implements the relations of Eqs. (1)-(3) is explained. Equation (1) is rearranged as

$$\theta(s)= (P(s)-A_f(s)+A_e(s))/ G_x(s) \qquad (4)$$

This relation is met by utilizing following control mechanism. Firstly, value of the right hand side of Eq.(4) is calculated by measuring torque P(t) and estimating isometric torques, A_f*and A_e*(t). Then, the value thus calculated is given the position control system as a desired value of the prosthetic hand; the angle of the prosthetic hand is regulated so as to coincide with this value. In other words, the position of the prosthesis $\theta(t)$ is driven only by the difference between the isometric torques, $A_f^*(t)-A_e^*(t)$, in the absence of the disturbance torque. When the disturbance torque P(t) is applied, the position of the prosthesis deviates softly by the amount of $P(s)/G_x(s)$, like a flexible rubber.

A block diagram of these control mechanism is shown in Fig. 2. There are two separate units, a myoelectric signal processing unit and a neuromuscular prosthesis. EMG processing consists of rectifying and smoothing, by which the isometric torque A_f or A_e is estimated; estimated values of the flexor and extensor muscles are denoted by A_f* and A_e*, respectively. An optimal smoothing filter for the control of the prosthesis is determined in Section 3. The neuromuscular prosthesis consists of a terminal device and a servo control unit which implements the relation of Eq.(4), or which simulates dynamic behaviors of the neuromuscular control system. A servo control system of prosthetic hand created is detailed in Chapter 5.

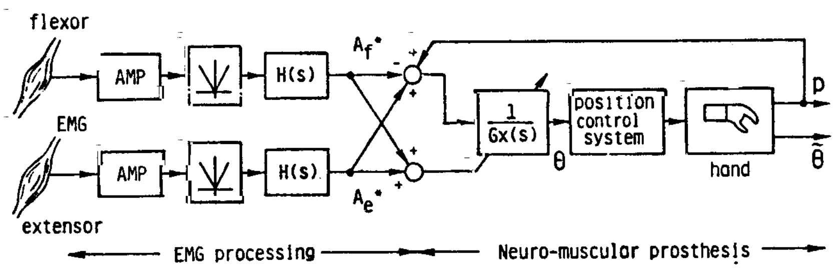

Fig. 2 Block diagram of the myoelectric prosthesis which is provided with dynamic properties of the neuromuscular control system, Eq.(1). A_f* and A_e* are estimated isometric torques of the flexor and the extensor muscle, respectively.

3. MYOELECTRIC SIGNAL PROCESSING

An optimal smoothing filter $H(s)$ is determined so as to meet two following conditions; (1) the output of the processing unit A_f^* (A_e^*) agrees closely with isometric torque exerted by the muscle, and (2) the subject can smoothly control movements of the prosthesis. This section begins with estimating the filter $H(s)$ from experimental data. Then, the optimal filter is determined, based on myoelectric control experiments of the prosthesis.

A model in Fig. 3 is used to represent the relation between EMG signal and isometric torque exerted by the muscle. The filter $H(jw)$ can be estimated by measuring both rectified EMG signal $r(t)$ and isometric torque $y(t)$;

$$H(jw) = G_{ry}(jw)/G_{rr}(w) \tag{5}$$

where $G_{rr}(w)$ is the auto-spectral density function of $r(t)$ and $G_{ry}(jw)$ the cross-spectral density function of $r(t)$ and $y(t)$.

The experiments were carried out both for extensor (extensor digitorium communis) and flexor (flexor carpi radialis and ulnaris) muscles in three healthy subjects. Surface EMG signals and torques around the metacarpophalangeal joint were simultaneously observed under the isometric condition. The subjects were instructed to change the torque almost triangularly at a certain frequency. The similar experiments were repeated eight times at various frequencies of 0.0625 to 2 Hz. Data length of one series of experiment was 8.2 sec. Frequency responses $H(jw)$ were calculated by using Eq.(5) and then eight responses were averaged. The responses were almost the same regardless of the subjects or the muscles.

In order to implement the smoothing filter with an analog circuit, the frequency response has to be expressed in a form of the transfer function. It was indicated that a second order transfer function approximately fits the frequency responses. The approximation, however, was not good enough because of nonlinearities. Therefore, three transfer functions were determined from three different criteria (fitting of impulse response, gain characteristics and gain.phase characteristics). Then, following experiment was carried out to determine an optimal filter for the prosthesis. Using the prosthesis described later in Section 5, the subject was asked to execute pursuit tracking tasks by means of EMG signals, where the reference input was sine wave of 0.5 Hz or 0.25 Hz and surface EMG signals of the extensor digitorius and the flexor carpi radialis muscle were picked up. Judging from the performance of pursuit tracking task (measured positional errors, smoothness of the movement and the subjects' opinion), following filter was determined as an optimal one;

$$H(s) = K_1/(s^2+2\ d\ w_n\ s\ +\ w_n^{\ 2}) \tag{6}$$

$$K_1 = 42\ \ (N/mV), \quad d = 0.85, \quad w_n = 14.5/\mathrm{sec.}$$

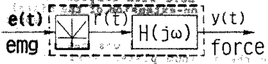

Fig. 3 Mathematical model representing the relation between EMG and isometric force.

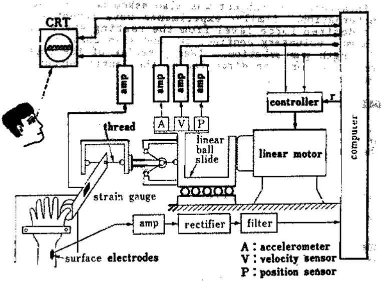

Fig. 4 Experimental apparatus for examining the myotatic reflex responses of the flexor pollicis longus muscle of the thumb.

4. ESTIMATION OF DYNAMICS OF THE NEUROMUSCULAR CONTROL SYSTEM

4.1 Methods

An experimental set up is schematically shown in Fig. 4. The experiments were carried out on the left flexor pollicis longus (FPL) muscle of three healthy subjects. The forearm was fixed horizontally and the proximal phalanx of the subject's hand was held with compound so that movement was restricted to flexion-extension of the distal joint. The tip of the thumb was connected to a DC linear motor by means of an alminum plate , dacron threads and linear ball slide. The torque exerted by the thumb was measured by strain gauges mounted on the alminum plate. Position was measured by a linear position sensor that monitored the position of the motor. Since a linear relation between the angle of the thumb and the measured position was confirmed, force and position measured at the tip of the alminum plate can be converted to torque and angle, respectively, by using the length of the lever arm (5 cm).

The subject was asked to maintain constant force while watching a strain gauge output displayed on an oscilloscope. Then, either ramp stretch or ramp shortening of approximately 5 mm (about 5.7 degree) was applied when the desired force was maintained. The subject was also asked not to react the length purturbation. Similar experiments were repeated with varying the desired force level from the resting state to 50 % of the maximum voluntary contraction, and also changing the direction of length purturbation. Twenty series of the experiment were performed for each direction of the length purturbation in one subject.

DATA ANALYSIS

Denoting mass of the thumb and the measuring instrument by m, and viscous coefficient around the distal joint by B, measured force F(t) is given as a function of the position of the thumb U(t);

$$F(s)=F_f(s)+(G_x(s)+m\,s^2+B\,s)\,U(s) \qquad (7)$$

where $F_f(s)$ is isometric force exerted by the FPL muscle and $G_x(s)$ is expressed by Eq.(2).

First step is to estimate the parameters m, B, K_o, T_1 and T_2 from the experimental data which were obtained by stretching the thumb at the resting state: the parameters were determined, by means of Simplex method, from the best least-squares fitting of the torque data, i.e., minimizing the squared error between the model output of Eq.(7) and the measured force. Secondly, parameters K, T_1 and T_2 of the contracting muscle were determined, by applying the same best least-squares fitting technique to the experimental data that were obtained from stretching the thumb exerting constant tonic force. Because voluntary reaction time of the subjects studied were approximately 150 ms, the experimental data used for the fitting were those over the interval from the onset of the stretch to 150 ms after the onset of stretch.

4.2 Results

One of the experimental records is shown in Fig. 5. First increase of the stretch-evoked force lasting up to about 50 ms after the onset of stretch is attributable to viscoelasticity of the muscle, and the second increase whose peak is around 100 ms after the stretch is attributable to the myotatic reflex.

The force response calculated from the model at the best fit is shown on the lower trace. Apparent difference is seen between the measured force and the model response. Closer agreement could be obtained if more complex and elaborated equation including time delay or nonlinearity was adopted, rather than Eqs.(1)-(3). If so, the hardware of the prosthesis would be more complex and heavier. Since the model represents essential dynamic properties of the neuromuscular control system of the finger muscle, Eqs.(1)-(3) are considered sufficient as a first approximation model.

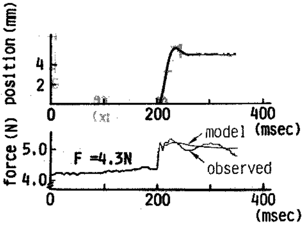

Fig. 5 Simulation result.
Observed force of the flexor pollicis longus muscle in response to stretch, and the calculated force of the model.

Parameters estimated from one subject are plotted against the tonic force in Fig. 6, where the length purturbation was stretch. It is showed that gain K was proportional to the tonic force while time constants T_1 and T_2 were not dependent on the tonic force. The same characteristics were obtained from all subjects studied and for both directions of the length purturbation. The parameter K_o and a are determined from the regression lines of K, and mean values of T_1 and T_2 are calculated. The parameters obtained from three subjects are given in Table I. Since neither estimated parameters varied with the subjects nor depended on direction of the length purturbation, averaged values given at the bottom in Table I were employed in implementing the prosthetic control system. Dimension of the gain K was converted from N/m of the linear movement to Nm/rad of the rotary movement of the prosthesis, using the lever arm of 5 cm.

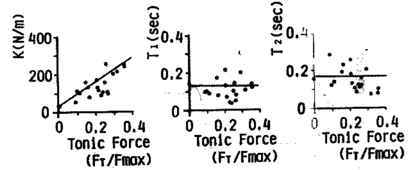

Fig. 6 Estimated parameters plotted against tonic forces of the FPL muscle. Fmax=maximum voluntary force, Ft=tonic force.

Table 1. Estimated Parameters of FPL Muscles.

	sub. (Fmax)	a (Nm/rad /100%MVC) (corr.cof)	T_1 (sec) mean S.D.	T_2 (sec) mean S.D.	K_0 (Nm/rad)
stretch	S.T (20.8N)	1.74 (0.724)	0.134±0.092	0.171±0.088	0.78×10^{-1}
	T.J (17.7N)	1.05 (0.442)	0.142±0.126	0.216±0.121	1.30×10^{-1}
	F.T (24.8N)	0.97 (0.779)	0.054±0.029	0.197±0.112	0.58×10^{-1}
shorten-ing	S.T (20.8N)	1.10 (0.608)	0.130±0.091	0.319±0.113	0.95×10^{-1}
	T.J (17.7N)	1.04 (0.593)	0.128±0.081	0.308±0.115	1.46×10^{-1}
	F.T (24.8N)	1.11 (0.545)	0.117±0.061	0.317±0.145	1.20×10^{-1}
	mean	1.3	0.12	0.25	1.0×10^{-1}

5. MECHANISM OF THE PROSTHETIC HAND

A one-degree-of-freedom prosthetic hand is created by way of trial for laboratory testing. Its configuration is shown in Fig. 7. Similarly to Otto Bock hand, the index and middle fingers are clamped together, and the thumb rotates in linkage with them by means of a linkage arm . The overall transmission ratio of 120 to 1 is implemented with a worm-gear unit; rotary motion is transmitted from DC motor (Minimotor 3540) to the fingers. The transmission system is designed so that the actuator is capable of producing the maximum pinch force of 5Kg at the tip of the finger and rotating the finger at the maximum speed of 200 degree/sec.

Strain gauges are attached on proximal surfaces of three fingers to pick up torque P; a linear sum of individual torques of the finger is obtained by weighted sum of the strain gauge outputs. For implementing the transfer function $1/G_x(s)$, a lead-lag filter $1+T_2s/(1+T_1s)$ is created with operational amplifiers and a variable gain $1/(K_o + a(A_f{*}+A_e{*}))$ with a devide circuit using an IC multiplier. A position control system employs a PD (proportional and differential) control for positional error, where output signal of an optical rotary encoder connected to the DC motor are fed back to the PD controller (phase-locked controller GL-1200, GAlil Motion Control) and a high efficiency , 25KHz, pulse-width-modulator is used for the motor control.

Frequency responses of the developed prosthesis were measured to examine the control performance; angle of the prosthetic hand was observed when an input $A_f{*}(t)-A_e{*}(t)$ changed sinusoidally by means of an oscillator and another input, $A_f{*}(t)+A_e{*}(t)$, was constant (gain K is fixed). A frequency characteristics measured is shown with solid line in Fig. 8, which closely agrees with the theoretical curve of $1/G_x(s)$ up to about 10 Hz. This characteristics is considered practically sufficient for the amputees to myoelectrically control the prosthesis.

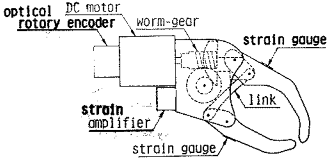

Fig. 7 Configuration of one-degree-of-freedom prosthetic hand.

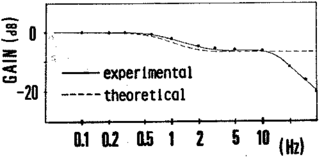

Fig. 8 Frequency response of the prosthesis between $A_f^*-A_e^*$ (input) and the angle of the prosthesis $\tilde{\theta}$ (output) ; $A_f^*+A_e^*$=constant, T_1=0.12 sec, T_2=0.25 sec.

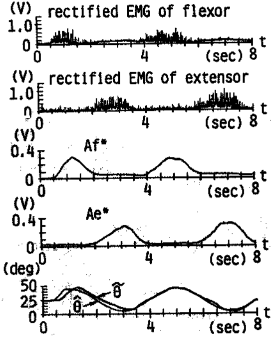

Fig. 9 Results of myoelectric control experiment of pursuit tracking; $\hat{\theta}$ is desired angle, and $\tilde{\theta}$ is the angle of the prosthesis.

6. MYOELECTRIC CONTROL EXPERIMENTS

To assess usefulness of the developed prosthetic hand, preliminary myoelectric control experiments were carried out with healthy subjects, by utilizing the smoothing filter determined in Section 3, the terminal device and control system described in Section 5. Surface EMG signals of extensor digitorium and flexor carpi ulnaris muscles of the right forearm were picked up. The sum of and the difference between rectified, smoothed EMG signals of two antagonists muscles were used. The subject was asked to coincide an angle of the prosthetic finger with the desired angle, watching both angles displayed on an oscilloscope. One of the typical results is shown in Fig. 9. It is seen that the angle of the prosthesis coincide closely with the desired angle . Note that only a half hour training was enough for the subject to perform these control experiments. It is thus expected that the prosthesis would be easy for the amputee subject to operate.

Following experiments were conducted to examine how softly the prosthetic fingers behaved in response to disturbance force and to what extent the prosthesis dynamics were adjusted according to activities of the muscles. External disturbance force was applied manually to the finger, when the angle of the prosthesis was kept almost constant, the subject being maintaining steady activities of two antagonist muscles. One of the experimental results is shown in Fig. 10(A). It is seen that the finger responded softly to the disturbance like a natural limb. Similar experiments were performed with varying activities of the muscles. The activities in Fig. 10(B) was lower than those in (A). Adjustments of the prosthesis dynamics is made clear by comparing two responses of (A) and (B). That is, when the activities were at the lower level, smaller disturbance force evoked almost the same angular deviation, as shown in Fig. (B). In other words, high levels of contraction of the antagonist muscles cause stiffering of the hand, while low contraction levels allow graceful, free-swinging behavior. These dynamic behaviors are just like those of a natural limb. It is apparent that these features of the prosthesis are accomplished by a variable gain K which is controlled by the sum of antagonist EMG signals, $A_f^*+A_e^*$. Recently, using Utah Arm, Jacobsen et al. have investigated effects of varying torque and speed feedback gains on elbow dynamics, and confirmed capability of adjusting elbow dynamics 2).

7. Summary

The present study attempted to provide the prosthetic hand with functions of the neuromucular control system which involved functions of two antagonist muscles and of the myotatic reflex, for the purpose of restoring partially the lost functions of the amputees. High performances of the prosthesis were showed by performing the myoelectric control experiments with the healthy subjects. Since the prosthesis created in the present study was only for laboratory testing, a lighter and more reliable prosthesis which employs digital circuits for minimizing power comsumption is currently under development.

The results obtained in the present study are summalized below.

(1) Fundamental dynamic properties of the neuromuscular control system were determined from finger muscles in man. Myoelectrically controlled prosthetic hand which had the dynamic properties thus determined was developed by utilizing the position control system and torque feedback.

(2) The sum of and the difference between processed EMG signals of two antagonist muscles were used as control command signals to the control system. The processing of EMG signals consisted of rectifying and smoothing, and an optimal smoothing filter was determined.

(3) The prosthetic hand was smoothly controlled myoelectrically by the healthy subjects after a brief training period. Because the mechanical impedance of the prosthesis varied with activities of the muscles, the finger of the prosthesis behaved softly like a natural limb ,in response to applied disturbance force.

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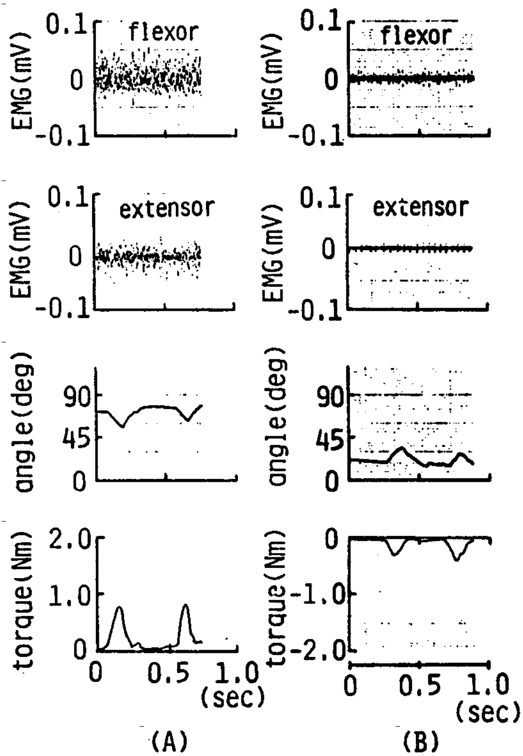

Fig. 10 Illustrating the variable mechanical empedance. Responses of the prosthesis to disturbance force applied to the prosthetic finger. Intensity of the contraction was at the high level in (A) and at the low level in (B).