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Compliant Grasp in a Myoelectric Hand Prosthesis

Controlling Flexion Angle and Compliance with Electromyogram Signals

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A novel myoelectric hand prosthesis consisting of electromyogram (EMG) signal processing units, a microprocessor-based dc motor servo system, and a 1 degree-of-freedom (DOF) end effector has been developed. The flexion angle and compliance of the finger of this prosthesis can be voluntarily controlled with EMG signals. Basic functions of the human neuromuscular control system are realized by using position control, force feedback, and variable gain, modulated by EMG signal amplitude. A limb-absent person and four healthy subjects were able to voluntarily control the finger angle and compliance of the prosthesis and were able to easily grasp a soft object after a short training period.

Powered hand prostheses are used to replace the functions of a lost natural hand. Most of the commercial prosthetic hands in clinical use are controlled by myoelectric signals (EMG unit) and are referred to as myoelectric hands [1]–[4]. This article describes the Osaka Hand, a biomimetic control system for a more conventional prosthetic hand mechanism. The advantages and shortcomings of each control system are described in the discussion.

Control systems for myoelectric hands can be classified roughly into two types: autonomous and voluntary. In an autonomous control system, when the user gives simple supervisory instructions to the prosthetic hand, the angle of each joint is controlled automatically. In the Southampton Hand [5], [6] and the MyoBock System Electric Hand, SensorHand [7], the grip force is readjusted automatically when slip is detected by slip sensors in the digits.

There are two types of voluntary control system: on-off (confusingly referred to as *digital*) and analog. In a digital control system, when the amplitude of the EMG signals exceed a determined threshold, the fingers open or close, and the velocity is fixed [8]–[12]. One of the best-known digital systems is the MyoBock System Electric Hand by Otto Bock Orthopedic Industry [8]. This system uses EMG signals detected from a pair of flexor and extensor muscles. Prostheses based on this system have been used extensively with children and adults in many countries. Another type of digital system used in the past was the one-site, three-state system, which was developed at the University of New Brunswick, Canada [9]. In this system, a single-channel EMG signal (one site) is used as the control signal. The three states are locked, closing, and opening; the hand is locked when the muscle is relaxed, closing when the muscle contracts weakly, and opening when the muscle contracts strongly. This control system was used as an alternative to the standard Hugh Steeper Hand control system [4].

In analog control systems, the rectified and smoothed EMG (RSEMG) signals are used as control signals. Two types of analog control systems have been reported: proportional [7], [13]–[15] and compliant [16], [17]. In a proportional control system, the velocity of the opening or closing of the hand increases with increasing difference in RSEMG between the flexor and extensor muscles. Recently, the MyoBock System Electric Hand DMC Plus has been established as a clinical product [7].

A compliance control system has been developed as part of a biomimetic hand prosthesis. This control system allows users to voluntarily control both the finger angle and the compliance. Skeletal muscles have well-known viscous and elastic properties [18], and the dynamic properties of both the muscle itself and the stretch reflex are not fixed but change depending on the activation level of muscle contraction [19]–[22]. These properties play an important role in maintaining posture and controlling limb movements as well as in absorbing mechanical impact exerted on the limb. A prosthetic hand that restores these lost functions should allow the user to easily grasp both soft and hard objects by controlling the compliance of the hand.

Based on these considerations, the biomimetic myoelectric hand is designed to mimic the basic functions of the neuromuscular control system of human finger muscles [16], [17]. In a previous study, it was found that two limb-absent subjects were able to voluntarily control the hand and, therefore, grasp soft objects [17]. In subjective evaluation, the subjects reported that the compliance control was useful when they grasped the soft objects. However, the power of the dc motor of the prosthesis did not provide sufficient control of compliance to allow practical use of the hand. The present study moves towards practical, clinical use. An improved version of the hand using a more powerful dc motor (6 W) and a recently developed advanced microprocessor unit (MPU) was evaluated. Once again, it was found that the users were able to voluntarily control the finger angle and the compliance and grasp soft objects.

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This article first describes the developments to the biomimetic hand then details the neuromuscular control system dynamics of a human finger that were used to design this biomimetic hand [16], [23]. The suitability of various parameters of EMG processing for use in the design of prosthetic hands [16], [24] are examined. A novel type of digital servo system is briefly described. Finally, an evaluation of the utility of the biomimetic hand is described. It was conducted with one limb-absent prosthesis user and four healthy subjects.

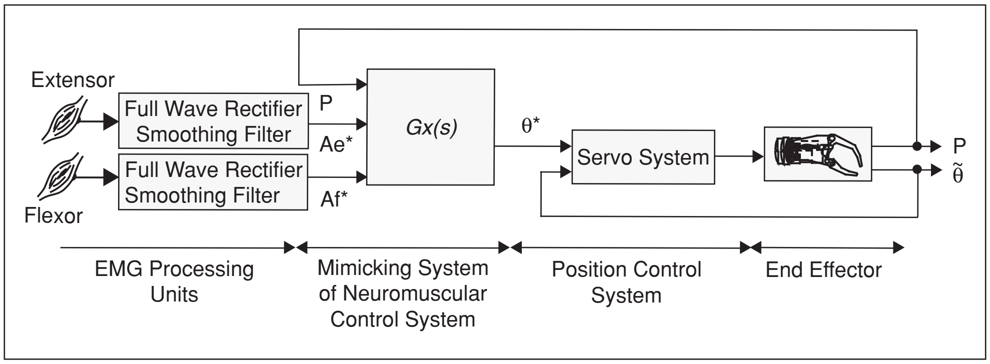

Fig. 1. A block diagram of the biomimetic hand. Angle and stiffness are controlled with EMG signals of flexor and extensor muscles.

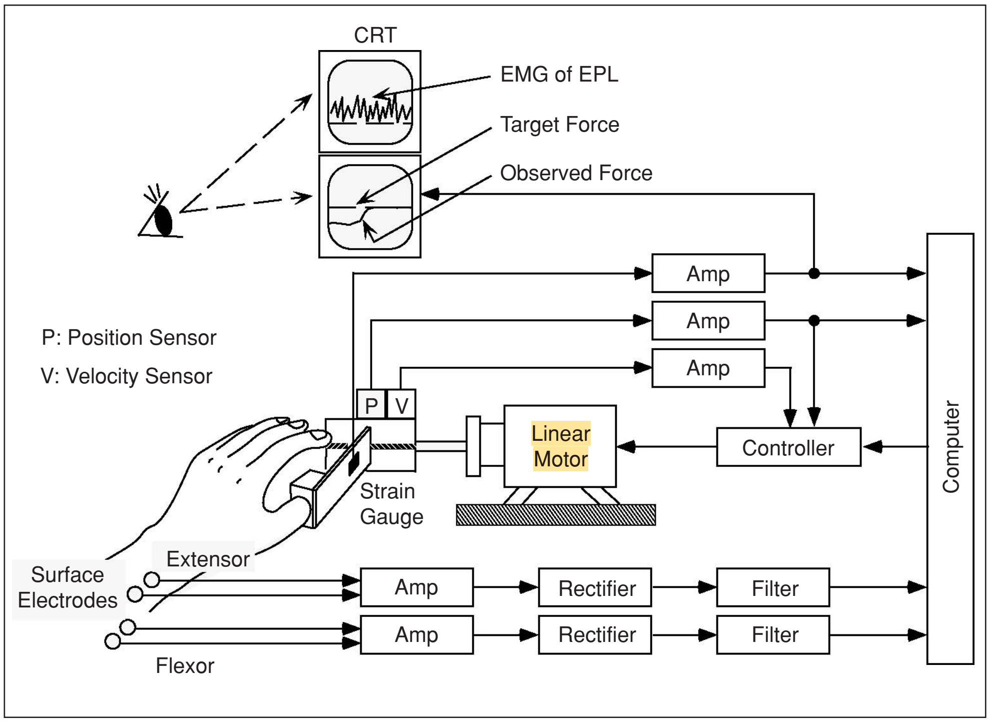

Fig. 2. Experimental setup: force and displacement are measured when length perturbation is applied to the FPL during contraction at constant isometric force.

Outline of Biomimetic Myoelectric Hand Controller

Figure 1 shows a block diagram of the biomimetic myoelectric hand. The hand consists of EMG processing units, a system to emulate the neuromuscular control system, a position control system, and an end effector.

Output of the EMG processing units is roughly related to the contractile force of the muscle. The outputs of the flexor and extensor are denoted by A_f^* and A_e^*, respectively. Details of this are described in the section "Design of Smoothing Filter of EMG Signals."

In order to emulate the neuromuscular control of the finger, the desired finger angle is calculated, based on simplified dynamics of a neuromuscular control system, which consists of the flexor and extensor muscles. The angle of the joint is denoted by $\theta(t)$ and the torque around the joint by $P(t)$. We define the joint output torque $P(t)$ as the difference between torque of the flexor $A_f(t)$ and torque of the extensor $A_e(t)$ when the joint is maintained at the angle $\theta(t) = 0$. Assuming that when the joint angle changes from $\theta(t) = 0$, torque is added depending on the muscle length; this implies that the torque is due to the viscous and elastic properties of the muscles and the stretch reflex. Denoting this relation by the transfer function $G_x(s)$, the dynamic equation of the neuromuscular control system is expressed as

$$P(s) = A_f(s) - A_e(s) + G_x(s)\theta(s). \tag{1}$$

Details of this formulation are described in the following section of this article. From this equation, we obtain

$$\theta(s) = \{A_f(s) + A_e(s) - P(s)\}/G_x(s). \tag{2}$$

Measuring the external torque P applied to the finger and the torque A_f^* and A_e^*, we obtain the desired angle θ^* as follows:

$$\theta^*(s) = \{A_f^*(s) + A_e^*(s) - P(s)\}/G_x(s). \tag{3}$$

If the position control system is perfectly accurate, the finger angle of the end effector $\tilde{\theta}$ is equal to the desired angle obtained using (3). In general, the position control system has the dynamics $G_{\text{motor}}(s)$, so the actual angle is expressed as

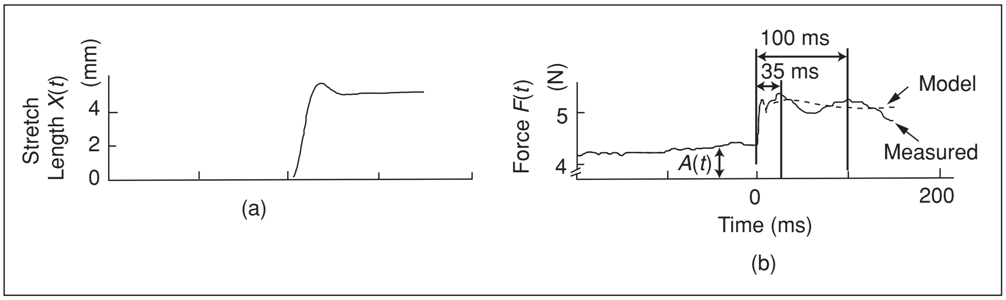

Fig. 3. The measured force of the FPL muscle in response to length perturbation (stretch, isometric contraction force, 20% maximum voluntary force (MVC), 4.3 N): (a) stretch length applied to the thumb $X(t)$ and (b) measured force response $F(t)$ (solid line) and calculated force response (broken line).

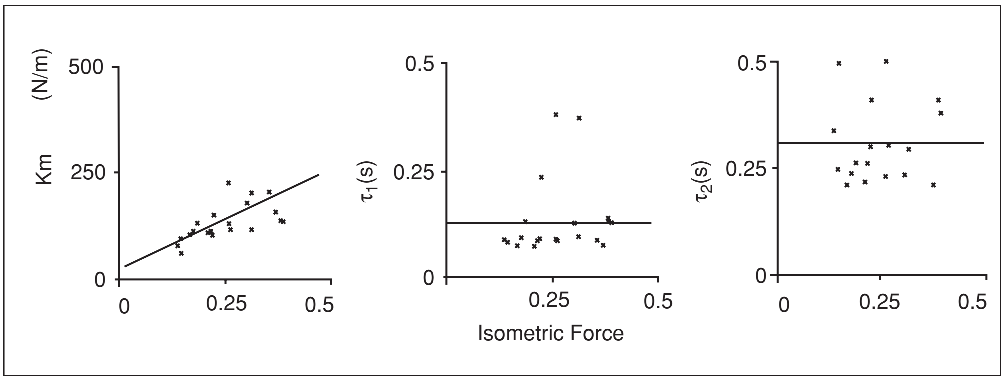

Fig. 4. The relation between estimated parameters of $G_x(s)$ and tonic force normalized by the maximum voluntary force. The tonic force, defined as the isometric force, is measured at onset of the length perturbation.

$$\tilde{\theta}(s) = G_{\text{motor}}(s)\theta^*(s). \qquad (4)$$

Modeling of the Human Neuromuscular Control System

To identify the dynamics of the human neuromuscular control system $G_x(s)$, the force response to a small-length perturbation of the human flexor pollicis longus (FPL) muscle was measured. The subjects were three healthy males who gave informed consent. The experimental setup is shown in Figure 2. The tip of the thumb was fixed to an aluminum plate, and the tip of the plate was connected to a linear ball-slide that was connected to a voice coil motor. The force exerted by the thumb was detected using strain gauges fixed to the aluminum plate. Displacement of the thumb was measured using a position sensor. Surface EMG signals from the FPL and extensor pollicis longus (EPL) muscles were detected. The signals were full wave-rectified and smoothed with a second-order low-pass filter (cutoff frequency: 2.6 Hz).

A ramp length perturbation of stretching and contracting (5 mm; duration: 17 ms) was applied to the thumb while the subject maintained constant isometric force. The subjects were asked not to react against the length perturbation. These measurements were repeated with variation of the isometric force levels from a resting state to 50% of the maximum voluntary contraction (MVC) and changes in the direction of the length perturbation. With each subject, 20 measurements were performed for each direction of the length perturbation.

Figure 3 shows one of the experimental records. The stretch length perturbation $X(t)$ is shown in Figure 3(a). The force response $F(t)$ is shown in Figure 3(b). The first peak around 35 ms was attributed to the viscouslike and elastic properties of the muscle, and the second peak around 100 ms to the long-latency stretch reflex.

Referring these responses to the length perturbation, a model of the neuromuscular control system was constructed, it is as simple as possible and applicable to the prosthetic hand. It is assumed that $F(t)$ is the summation of $A(t)$ and $F_v(t)$, where $A(t)$ is the force that does not depend on the muscle length but rather depends on the muscle activation, and that $F_v(t)$ is the force evoked by the length change $X(t)$, which is mainly attributed to the intrinsic viscouslike and elastic properties of the contracting muscle and the stretch reflex. Although the dynamics of this system are complicated by nonlinearities and time delays, it can be expressed using a transfer function $G_v(s)$.

$$G_v(s) = F_v(s)/X(s) \qquad (4)$$

$$F(s) = A(s) + G_v(s)X(s) \qquad (5)$$

As the first step, we express $G_v(s)$ in the simplest form

$$G_v(S) = K_m \frac{1+\tau_1 s}{1+\tau_2 s}, \qquad (6)$$

where τ_1 and τ_2 are time constants.

The parameters τ_1, τ_2, and K_m were estimated by obtaining the best fit of the model response against the observed force response over the period of 150 ms after the onset of length perturbation. One of the simulation results is shown in Figure 3. Although the model response did not sufficiently coincide with the measured force, it was concluded that the basic dynamics of the finger muscle can be represented by the model as a first approximation.

The estimated parameters are plotted against the isometric force in Figure 4. The results show that gain (stiffness) K_m increased almost linearly with the tonic force (the isometric force

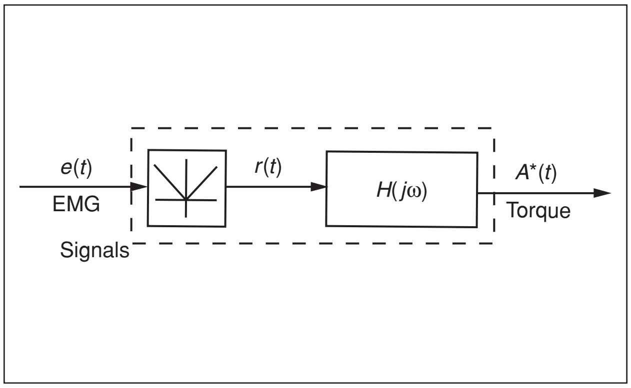

Fig. 5. The configuration of the EMG processing unit.

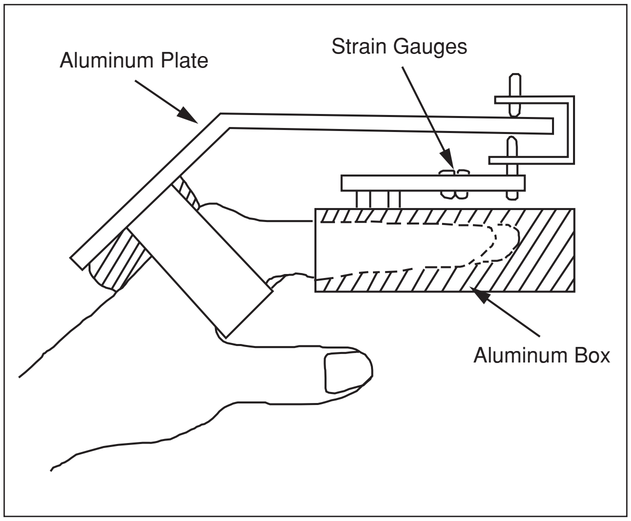

Fig. 6. An experimental setup for the design of the smoothing filter.

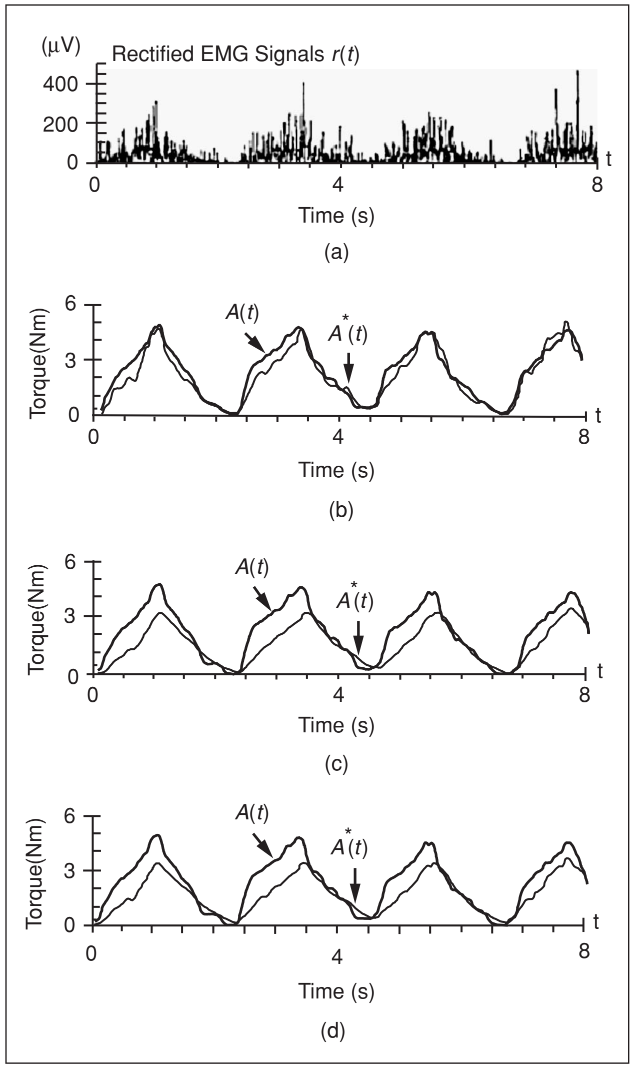

Fig. 7. Rectified EMG signals of the extensor muscle and isometric force. Thin lines, $A(t)$, are measured force. The thick lines in (b), (c), and (d) are force curves, $A^*(t)$, obtained using the smoothing filters A, B, and C, respectively.

measured before the length perturbation); the force is normalized by MVC force. The time constants τ_1 and τ_2 do not appear to be dependent on the force, although exact values of the time constants could not be determined because of the low sensitivity of the estimation of these time constants. The same tendencies were observed for all subjects and for both directions of the length perturbation. Mean values of the time constants and the slope of K_m against the tonic force were calculated: $\tau_1 = 0.12$ s, $\tau_2 = 0.25$ s, and gain (stiffness) at MVC $(K_m) = 15.5$ Nm/rad, which was converted from the value of the linear system to that of the rotational system. The dynamics of the muscle at the resting state were also estimated. Finally, the dynamics $G_x(s)$ in (1), were expressed as follows:

$$G_x(S) = K\frac{1+\tau_1 s}{1+\tau_2 s} \tag{7}$$

$$K = K_0 + a(A_f + A_e). \tag{8}$$

Note that the gain (stiffness) K is time variant and changes proportionally with the sum of the exerted contractile force (torque) of the antagonistic muscles, $A_f + A_e$. The values $K_0 = 0.1$ Nm/rad, $\alpha = 0.98$ /rad, $\tau_1 = 0.12$ s, and $\tau_2 = 0.25$ s were used for human finger muscles, where K_0 represents the resting state [23].

Design of Smoothing Filter of EMG Signals

The model in Figure 5 expresses the contractile force $A^*(t)$ as a function of EMG signal $e(t)$, where $r(t)$ is the output of a full wave rectifier and $H(j\omega)$ is the output of a smoothing filter. A parametric filter $H(j\omega)$ was needed for the prosthetic hand.

The smoothing filter was designed as follows. First, the subject was asked to perform voluntary isometric contraction to measure EMG signals $e(t)$ and the isometric torque $A(t)$. Then, the filter $H(j\omega)$ was calculated as follows:

$$H(j\omega) = G_{rA}(j\omega)/G_{rr}(j\omega), \tag{9}$$

where $G_{rr}(j\omega)$ is the autospectral density function of $r(t)$, and $G_{rA}(j\omega)$ is the cross-spectral density function of $r(t)$ and $A(t)$. Next, transfer functions of the smoothing filter $H(s)$ were calculated from different criteria. Finally, an optimal transfer function was selected from among these transfer functions by examining performance of the finger angle control.

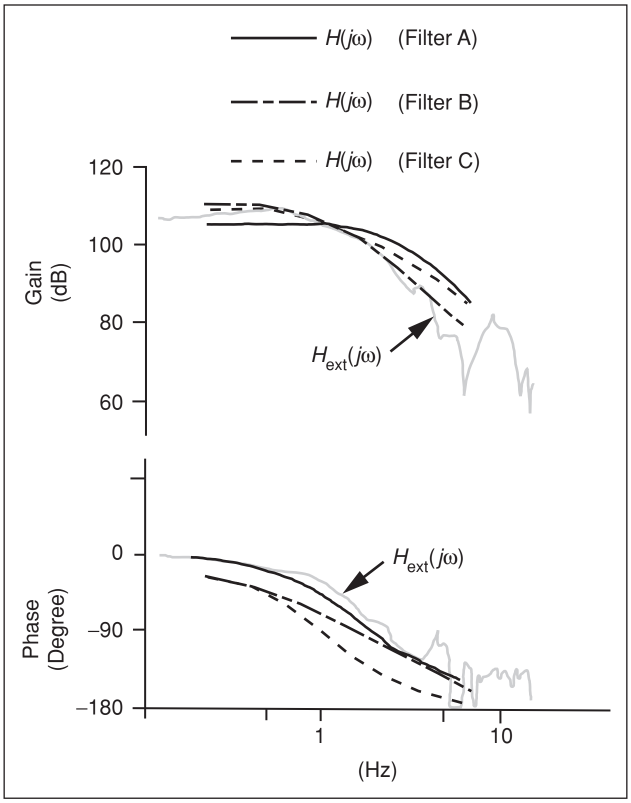

Fig. 8. Frequency characteristics. Thin lines indicated by $H_{ext}(j\omega)$ are experimentally obtained frequency responses. Thick lines are frequency responses of the transfer functions obtained using the approximation methods A, B, and C.

The experiment was conducted with three healthy subjects using the experimental setup shown in Figure 6. Four fingers were fixed in an aluminum box that was fixed to an aluminum plate. The force (torque) was detected with strain gauges, as shown in Figure 6. Surface EMG signals from the extensor digitorium communis muscle and torques around the metacarpophalangeal joint were simultaneously recorded under isometric conditions. The subjects were asked to change the torque almost triangularly at a certain frequency; one of the results is shown in Figure 7. The experiments were repeated eight times at various frequencies between 0.0625–2 Hz. The data length of each experiment was 8.2 s.

The calculated frequency response $H_{ext}(j\omega)$ is shown in Figure 8. It is the average of eight responses. The frequency responses were approximated with second order transfer functions according to three different criteria: a) fitting of impulse response, b) fitting of gain characteristic, and c) fitting of both gain and phase characteristics. Frequency characteristics of three transfer functions are shown in Figure 8.

In order to choose the transfer function of the optimal filter for controlling the prosthetic hand, performance with voluntary control experiments was evaluated. The subject was asked to make the angle coincide with the indicated angle (sinusoidal signal), with both angles displayed on a CRT and the prosthetic hand represented by (3), (7), and (8). Based on the error, we selected Filter A as the optimal filter, using the following equation:

$$H(j\omega) = \frac{K_n \omega_n}{s^2 + 2\zeta\omega_n s + \omega_n^2}, \tag{10}$$

where $K_n = 30.1 Nm/mV,\ \omega_n = 15.5/s,\ \zeta\omega_n = 13.2/s.$

Configuration of the Digital Servo System of the Biomimetic Hand

Figure 9 shows the configuration of the biomimetic hand developed. It consisted of a socket, two pairs of surface electrodes, a digital servo system, and an end effector. The sockets were constructed to fit over the forearms of the subjects. The end effector was attached at the distal end of the socket and consisted of a thumb, index finger, and middle finger, which

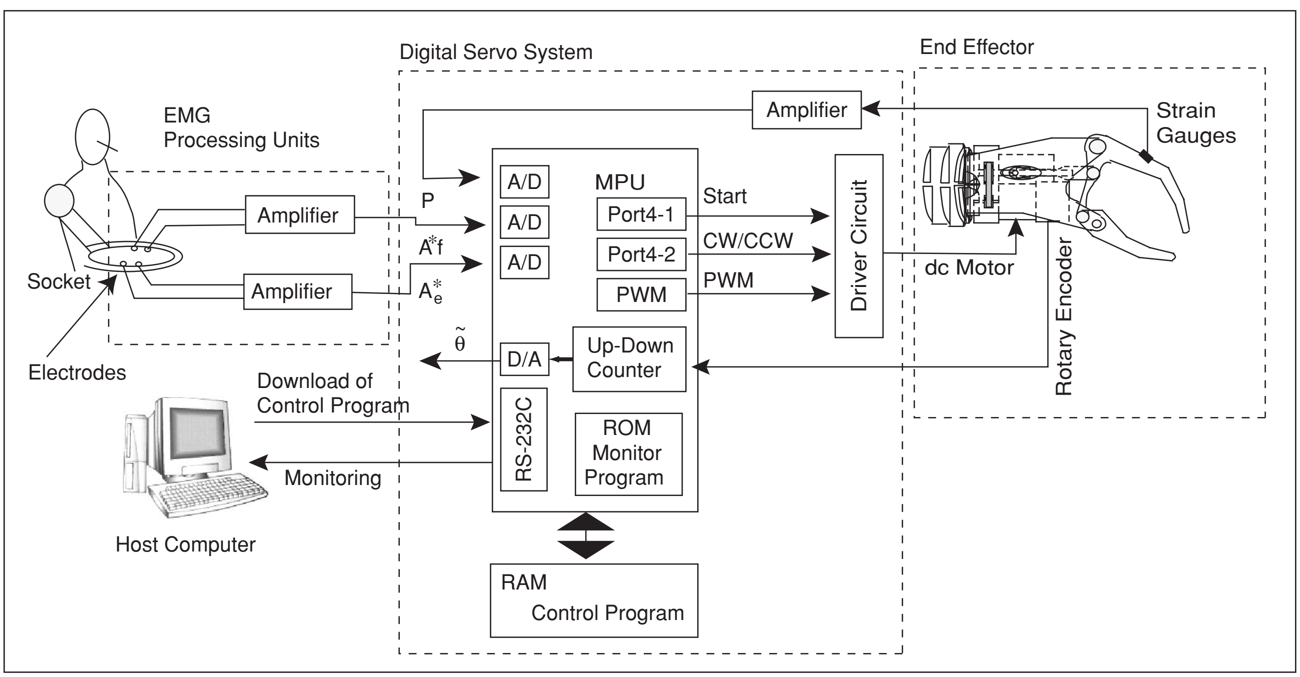

Fig. 9. The microprocessor-based digital servo system and computer system used to develop the control program of the prosthesis.

opened or closed simultaneously. The torque $P(s)$ applied to the fingers was measured with strain gauges attached to each finger. The surface EMG electrode was a dry type with an impedance converter. The electrodes were positioned inside the socket. The EMG signals were recorded from extensor carpi radialis brevis and flexor carpi radialis muscles.

Both the neuromuscular control emulator and the servo controller—proportional and differential (PD) controller—of the position control system in Figure 1 were installed in the digital servo system in Figure 9. We selected an MPU (H8-3067/F, Hitachi, that provides fast calculation and real-time control and selected a high power dc motor (A-max 22 mm, 6 W, Maxon) that provides fast mechanical response. The features of the MPU include A/D converters (10 b, 8 channel), D/A converters (8 b, 2 channel), 16-b up-down counters (2 channel), PWM signal output ports (20 kHz, 2 channel), serial communication ports (RS-232C, 2 channel), and an E^2PROM (128 kB) to store the monitor program. The control program was developed on the host computer (PC) and transferred to the external RAM via the serial communication port.

The power of the dc motor used in the present study was almost 1.7 times greater than that of the dc motor used in our previous study (Type 2233-012S, 3.7 W, Minimotor), and the weight of the present motor (54 g) was less than that of the previous one (61 g). The current consumption of the present dc motor is about 20% greater than that of the previous one.

In the MPU, the desired finger angle, $\theta^*(\mathrm{t})$ in (3) was calculated from the applied torque $P(t)$ and the torque $A_f^*(\mathrm{t})$ and $A_e^*(\mathrm{t})$ estimated from EMG signals.

The servo controller was used for position control, and the rotation angle and angular velocity of the dc motor were calculated from the information sent from the optical rotary encoder of the dc motor. PWM signals modulated by $\theta^*(\mathrm{t})$ and the signal used to determine the direction of the motor rotation were supplied to the dc motor drive circuit.

The frequency characteristics of the total digital servo system were assessed by changing the torque P (input) and observing the angle $\tilde{\theta}(t)$ with fixed values of A_f^* and A_e^*. Note that in this case, K was constant, so that the system behaved as a linear system. The frequency characteristics of the system were similar to those of the damped second-order system. The cutoff frequency was 3 Hz, which is much higher than the 1 Hz used in the previous study [17].

Myoelectric Control Experiments

Myoelectric control experiments were conducted with the prosthetic hand developed in the present study to determine whether the users were able to voluntarily control the finger angle and compliance and grasp soft objects. The finger angle and compliance control experiments were conducted with four healthy subjects. The object-grasping experiments were conducted with a limb-absent subject, who gave informed consent.

Finger Angle Control

The purpose of this experiment was to determine whether the subjects could control the prosthetic hand easily. First, the subjects were instructed to practice controlling the finger angle freely for 5 min. Then they were asked to track a step change in angle. Both the finger angle of the prosthetic hand and the target angle were displayed as bright horizontal lines on an oscilloscope. The initial angle was 40°, and the target angles were 10°, 20°, 30°, 50°, 60°, and 70°. Each experiment was performed as a series of three trials.

Figure 10(a) shows the results of the first trial, in which fairly good tracking was obtained. The subject took almost 3 s to reach the target, and there was some fluctuation in the achieved angle. Figure 10(b) shows the results of the third trial after a total training period of about 15 min. The subject took 1 s to reach the target. Muscular activity of the extensor was less at the third trial than at the first trial. The subject was able to control the finger angle and coarsely track the desired finger angle after training of about 15 min.

Compliance Control

In a previous study [17], it was qualitatively demonstrated that it was possible to control the compliance with that system by achieving transient angle response to an applied force. The purpose of the present static compliance experiment was to quantitatively determine with the present system whether compliance could be maintained at a constant value when muscle activity was maintained at a constant value and whether compliance varied with muscle activity. The subject was asked to cocontract his muscles without changing the finger angle of the prosthesis, i.e., he was asked to maintain activity of both muscles at 20% MVC by watching the RSEMG signals. This resulted in maintenance of a constant finger angle of the prosthetic hand. While the subject maintained 20% MVC, the weight was slowly hung from the thumb, which caused the finger angle to change; this was measured. The measurements were repeated three times. These experiments were conducted with several different weights and also at the resting state.

Figure 11 shows the relationship between weight and observed change of the finger angle; the slope of weight-angle relation corresponds to the compliance. An almost straight

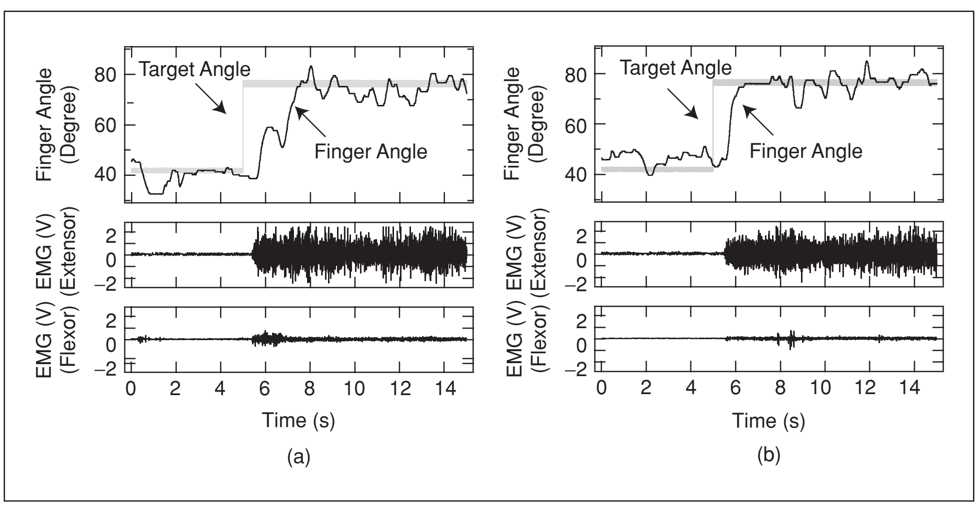

Fig. 10. Control of finger joint angle: (a) the first trial after training and (b) the third trial; muscle activity was less than in (a).

line was found, which implies that the prosthetic hand has a spring-like property, and almost the same compliance was maintained when the muscle activity was maintained at a constant value. The compliance was greater at the relaxed state than during contraction. These results are similar to what would be obtained with a normal human hand [21].

Object Grasping

In this experiment, a user of a conventional prosthesis grasped a soft object (a cream puff) with the prosthetic hand. Both of the subject's hands had been amputated below the elbow five years ago, and he has been using a body powered split hook; previously, he used a MyoBock System Electric Hand (on-off type). In an earlier laboratory study, he participated in experiments using another prosthetic hand we had developed.

First, the subject participated in preparatory testing of the control of the prosthetic hand (finger angle) for about 10 min. Then, the object-grasping experiment was conducted. The object was placed on a desk, and the subject was asked to grasp it, lift it up, and place it back on the desk. The experimental setup is shown in Figure 12.

Time profiles of the experiments are shown in Figure 13. At the first stage, the finger was opened to about 70° and was then closed on the object in about 1 s (indicated by an arrow).

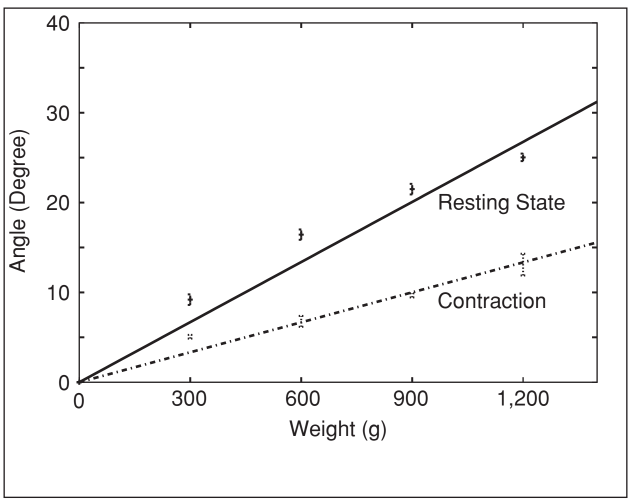

Fig. 11. The relation between weight and angle displacement at the resting state and at a steady contraction of 20% MVC.

Fig. 12. A photograph of the handling of a soft object with the prosthesis.

The subject grasped the object at Time A and released it at Time B. The results indicated that the subject could grasp the soft object easily and smoothly with the prosthetic hand.

The repeatability and reliability of the position of the surface electrodes were assessed after the initial grasping experiments. The subject removed the socket and then replaced it. Then the subject grasped the soft object, which was on the desk. These experiments were performed as a series of three trials. The subject was able to grasp the object without special adjustment, i.e., there was no need to readjust the parameters of the EMG processing units.

Discussion

Many kinds of voluntarily controlled myoelectric hands have been developed. However, none of them have been able to reproduce all of the important functions of the human hand. This is due to the lack of a compact and powerful actuator whose output power-to-weight ratio is the same as that of muscle and the inability to precisely detect and process a sufficiently large number of efferent and afferent signals to allow easy control of a multiple-DOF prosthetic hand. Thus, most prosthetic hands have limited functional range and are suitable for a limited range of tasks.

As mentioned previously, there are three main types of prosthetic hand control systems: on-off, proportional, and compliance control systems. The advantage of an on-off system is that it is simple, and the user can easily maintain constant finger angle. The disadvantage is that when grasping an object, the user must carefully watch both the object and the hand.

With a proportional control system, the user can easily control the speed of the finger movements. A proportional control system shares the previously described disadvantages. Furthermore, proportional control tends to require more current, and the battery life is generally shorter than that of an on-off system.

The advantage of a compliance control system is that the user can control finger angle and compliance, enabling the user to easily grasp hard and soft objects. Also, the user can determine finger angle without watching the prosthetic hand by sensing the contraction level of the muscles with proprioceptors. A disadvantage of such a system is muscle fatigue, resulting from the need to maintain muscle contraction to maintain opening or closing of the hand, without a similar shortened battery life (found to be approximately eight hours).

The overwhelming majority of prosthetic hands used in clinical settings is controlled completely voluntarily, whereas a natural human hand is partly autonomous. Some authors have discussed the need to include an autonomous system in the control system of a prosthetic hand. The advantage of an autonomous control system is that it can enable the user to easily control a multiple-DOF hand. However, the limited number of control commands available in an autonomous system limits the number of movement patterns that can be achieved. The ideal control system is one in which the control mode can be switched between autonomous and voluntary according to the situation. The Southampton Hand [5], [6] and SensorHand [7] include autonomous control systems. Users of these hands can control the finger angle voluntarily. However, when an object is grasped, the gripping force is readjusted automatically by detecting the slip of the grasped object using slip sensors attached to the fingers. Unlike the Otto Bock hand, the Southampton Hand can switch easily between autonomous and voluntary. Secondly, it uses additional information concerning the shape of the object to decide on the correct shape for prehension and to overcome the lack of possible inputs from the operator.

The aim was to develop a hand that emulates the functions of both the voluntary and autonomous control systems of the human hand. It appears that an autonomous system such as the one used in the Southampton Hand could be used in this prosthetic hand to vary the compliance. We would also include a system for transmitting information about slips and changes in the grasping forces to the user.

There have been many studies about pattern recognition of EMG signals [25]–[30]. Pattern recognition methods may be useful for controlling a multiple-DOF hand with an autonomous control system, with patterns of EMG signals corresponding to movements of the hand. It is difficult to detect and interpret the control signals necessary for the voluntary control of each joint of the hand. In an autonomous control system using pattern recognition, there is a limited number of possible movements of the hand. If pattern recognition can be made to work in real time, the user can operate the hand voluntarily. However, much research is needed before such methods can be applied to clinically useful prosthetic hands.

Some researchers have proposed using EMG signal processing for analog control of prosthetic hands [2], [31]–[33]. Hogan et al. have developed a method for deriving an estimate of muscle force [32], and Parker et al. have proposed a similar method [2]. Evans et al. have developed methods for generating control signals for a prosthetic hand using a full wave rectifier and a Kalman filter based on logarithmic nonlinearity [33]. Meek et al. have proposed a method for estimating the force of a muscle using an adaptive filter that varies its time constant according to the rate of change of the signal mean [31]. These processing methods require high-performance microprocessors, which generally require high current consumption. The prosthetic hand detailed here uses a smoothing filter with operational amplifiers of low current consumption.

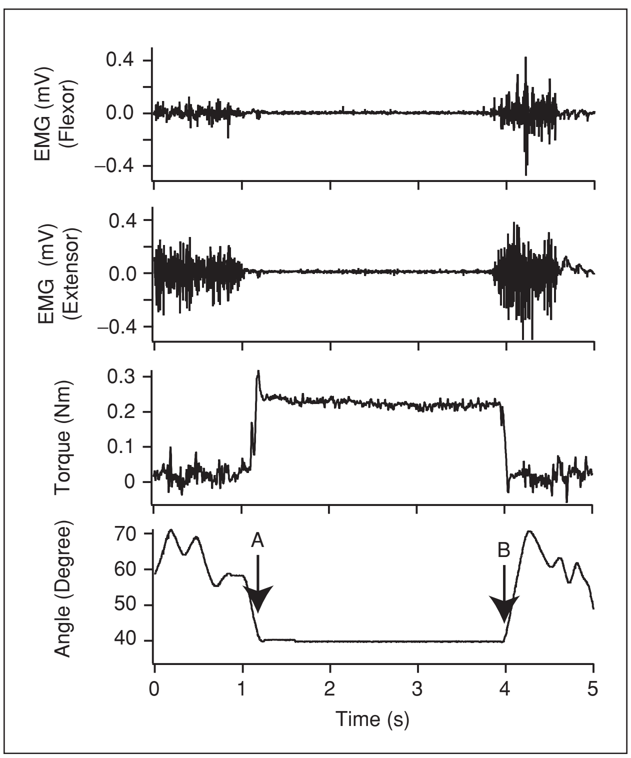

Fig. 13. Time profiles of the handling of a soft object shown in Figure 12.

In the present study, the values estimated from ramp stretch experiments with the FPL muscle were applied to the biomimetic hand. The results indicate that these parameters were useful for finger angle control, i.e., the four healthy subjects were able to voluntarily control the finger angle. However, it was difficult for the user to grasp the objects. This is because the desired dynamic range of the compliance could not be achieved, despite the greater power of the present dc motor. Therefore, the dynamic range of the compliance was adjusted (by decreasing the lowest level of compliance several times) so that the user could grasp both the hard and soft object after training in preliminary experiments.

The main feature of the present system is that the user can choose the parameter values of the prosthetic hand, e.g., gain of the EMG signal and coefficients of the dynamics of the neuromuscular control system that are most suitable for voluntary control of the hand.

Several issues must still be addressed before the present prosthetic hand can be used in daily life. For example, it requires a cosmetic glove that is flexible and tough. Also, it requires a supplementary sensory feedback system for practical use [34]. The user of this prosthetic hand must carefully watch the hand to determine the finger angle and pinch force of the hand. Although this prosthetic hand is useful in the laboratory, no attempt has yet been made to use it in daily life. Further study is required to determine whether the user of a conventional prosthesis could use the hand for a long period. This would demonstrate its usefulness and resolve any issues that may arise from daily use.

Conclusion

In this article, a biomimetic myoelectric hand was described, and a controller that allows voluntary control of compliance was demonstrated.

- ➤ The outline of the biomimetic controller for the myoelectric hand was described. It incorporated a model of the neuromuscular control system constructed from an analysis of the force response to length perturbation of the FPL muscle, processing of EMG signals, and the configuration of the hand.
- ➤ A novel digital servo system has been developed. It has been shown that the user can voluntarily control the finger angle and compliance and that an amputee subject can easily grasp a soft object after a short period of training.
- ➤ The features of the compliance control system of the biomimetic hand were described, and the advantages and disadvantages of the control systems used in conventional myoelectric hands were discussed.

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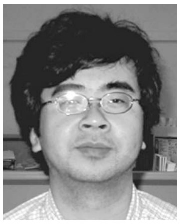

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