Control of IPMC-based Artificial Muscle for Myoelectric Hand Prosthesis

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Abstract: This paper proposes an ionic polymer metal composite (IPMC) based artificial muscle to be applicable to the myoelectric hand prosthesis. The IPMC consists of a thin polymer membrane with metal electrodes plated chemically on both faces, and it is widely applying to the artificial muscle because it is driven by relatively low input voltage. The control commands for the IPMC-based artificial muscle is given by electromyographic (EMG) signals obtained from human forearm. By an intended contraction of the human flexor carpi ulnaris and extensor carpi ulnaris muscles, we investigated the actuation behavior of the IPMC-based artificial muscle. To obtain higher actuation force of the IPMC, the single layered as thick as 800 [μm] or multi-layered IPMC of which each layer can be as thick as 178 [μm] are prepared. As a result, the bending force was up to the maximum 12[gf] from 1[gf] by actuating the single layered IPMC with 178 [μm], but the bending displacement was reduced to 6[mm] from 30[mm]. The experimental results using an implemented IPMC control system show a possibility and a usability of the bio-mimetic artificial muscle.

Key words: Artificial Muscle, Ionic polymer metal composite (IPMC), Bio-mimetic, Electromyography (EMG)

INTRODUCTION

Recently the rehabilitation engineering to restore physical abilities is regarded as an important technology for improving the quality of life. As the silver generation has been exponentially increasing, the social demands for the assistive devices such as prosthetic and orthotic devices to assist the muscle power are also increasing linearly. Most assistive devices use an electro-mechanical system like the electric motor for the assist force. However such power-assist system is heavy and not compatible with the human muscular system. Therefore artificial muscles to be implantable in human body are needed for the bio-mimetic power assist system.

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The electro-active polymer (EAP) was proposed as an artificial muscle to substitute for the human muscle [1]. Especially, ionic polymer metal composite (IPMC) is widely applying to the artificial muscle because it is simply manufactured and is driven by relatively low input voltage [2, 3]. The IPMC bends to the anode side when subjected to a voltage across its thickness as a result of cation migration toward cathode in the polymer network, because the cathode side of the composite swells and the anode side of it shrinks by the electro-osmotic drag of water induced by the cation migration [4]. The IPMC usually consists of a thin polymer membrane with metal electrodes plated chemically on both faces. Therefore, the IPMC has a wide usability in the soft robotic actuator and the dynamic sensors, because it can be utilized as an actuator module [1, 2, 5].

This study proposes an IPMC actuating system with a biomimetic function. The IPMC module is actuated by the human EMG signals generated by an intended muscle contraction. In this study, two muscles in forearm are used for the actuation of the IPMC. The bending force of the IPMC module is an important factor to be used for an artificial muscle. However, in [14] the bending force was less than 1 [gf]. In this

study, to obtain higher actuation force of the IPMC, the single layered as thick as 800 [μm] or multi-layered IPMC (Nafion) of which each layer can be as thick as 178 [μm] were prepared. To preserve the water existing within the IPMC for longer life performance, the each IPMC layer was coated and protected on the surface with some selected coating materials.

To investigate the IPMC actuation behaviors controlled with EMG signals [6, 7] generated from the human flexor carpi ulnaris and extensor carpi ulnaris muscles, we implemented an IPMC control system. The experimental results showed a possibility and a usability of the bio-mimetic artificial muscle.

EMG SIGNAL

EMG signal is the measured electric potentials produced by voluntary contraction of muscle fiber.

The frequency range of the EMG signal is within 0 to 2,000 [Hz] but the dominant energy is concentrated in the 30 to 500 [Hz] ranges, and its amplitude is limited to 0 to 10 [mV] (peak-to-peak) according to the muscle contraction [8]. Consequently, the muscle activation can be detected by measuring the amplitude of the EMG signal, because it is generally known that the amplitude of the EMG signal increases according to the increase of the muscle contraction force [7]. In this study, we apply the *mean absolute value* (MAV) of the surface EMG signal to the detection of the muscle activation. The MAV can be obtained by calculation from the digitized EMG signal. To implement a real-time control application, we have to reduce computational overheads. Thus, we use an active surface EMG sensor of which output is the MAV signal from the measured EMG signal.

Fig. 1 shows the block diagram of the active surface EMG sensor. The electric potentials obtained from two input electrodes are first amplified. The differential amplifier includes a band pass filter (BPF) with 4 [Hz] to 900 [Hz] bandwidths to select the EMG signal only, and its output signal is the raw EMG signal. However, 60 [Hz] noise due to the AC power line is mixed in the raw EMG signal.
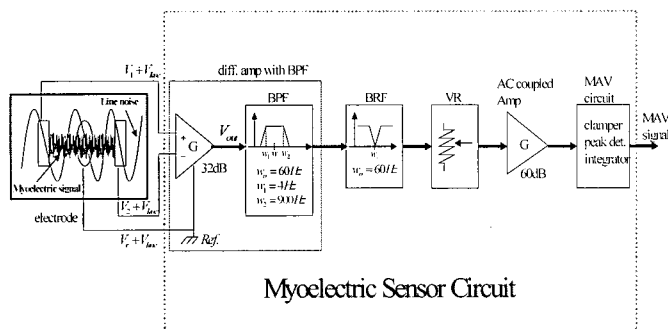
Therefore, we apply the band rejection filter (BRF) to the raw EMG signal. we then amplify the output of the BRF using AC-coupled amplifier with 60 [dB] gains. To adjust the difference of the source signal due to the muscle condition, we make a variable resistor in the input of the AC-coupled amplifier. Using a MAV circuit designed by a clamp and a peak detector circuit, the MAV signal with ripple is generated. Then, the final output is smoothed by a LPF with 1 [kHz] cutoff frequency.

IPMC

Actuation Principle of IPMC

The IPMC is composed of a perfluorinated ionic polymer, which is chemically surface composite with a conductive medium such as platinum (see Fig. 2). A platinum layer is formed a few microns deep within the perfluorinated ionic polymer. Typically, the strip of the perfluorinated ionic polymer membrane bends toward the anode (in the case of cation exchange membranes) under the influence of an electric potential. Also, the appearance of water on the surface of the expansion side and the disappearance of water on the surface of the contraction side are common. This electrophoresis-like internal ion–water movement is responsible for creating effective strains for actuation. Water leakage through the porous Pt electrode reduces the electromechanical conversion efficiency [2].
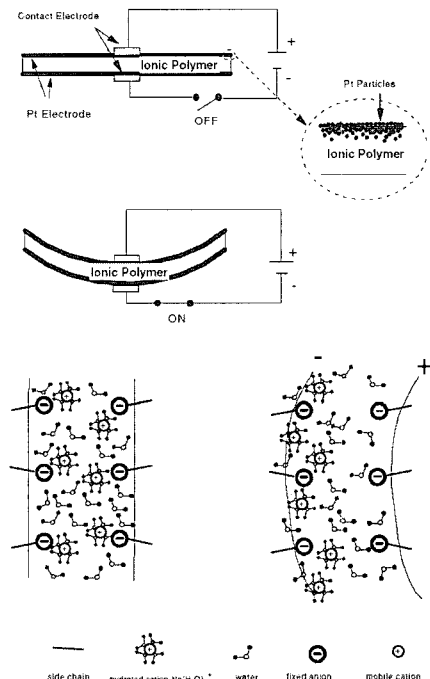

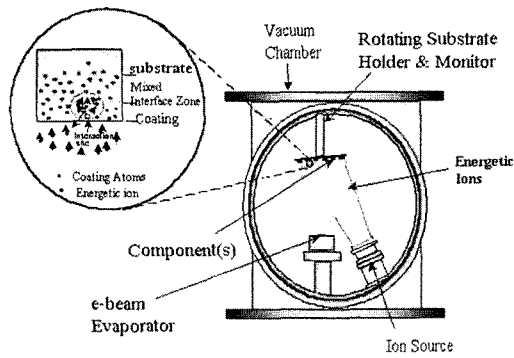Fig. 2. Schematic diagram of a typical IPMC and its actuation principle.

Fig. 1. Block diagram of EMG electrode

Since the passive filter attenuates the source signal,

Fig. 3. Schematic diagram of IBAD.

Preparation of IPMC

The actuation efficiency of IPMC depends on various factors including types of polymer membrane and electrode, the type of cation, the plating condition, the morphology and conductivity of the metal electrodes, surface roughening treatment, and the level of hydration [9]. As a very important factor among them, the electric property of surface electrode is mainly influenced by the molecular configuration of surface and membrane-metal interface.

In this study, the IPMC was prepared by adsorption-reduction method using Nafion 117 (H⁺-form, EW 1100, Dupont) and solution of platinum complex ([Pt(NH3)4Cl2], Aldrich Co.) [3]. The Ion Beam Assisted Deposition (IBAD) was adopted for surface electrode fabrication (see Fig. 3). After IPMC specimen set up in cyclic holder, the deposition of novel metals (Pt, Ir, and Au) was conducted on the surface of IPMC by electric gun (Telemark, USA) at acceleration voltage of 8.5 [kV] and deposition rate of 0.1 [□/sec], sustaining in vacuum of ~5×10⁻⁶ [Torr]. Figure 3 shows the schematic diagram of IBAD.
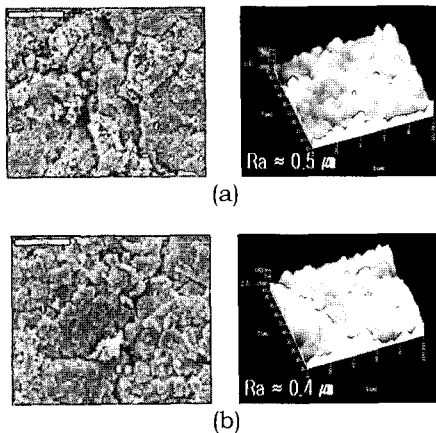

(a)

(b)

Fig. 4. Scanning electron microscopy (SEM) morphology and 3D-profiler of the IPMCs plated by (a) chemical reduction and by (b) physical IBAD (white bar = 10 [μm]).

The sandblasted Nafion was made into IPMC by adsorption-reduction method as initial compositing, and then surface electroding was carried out by conventional chemical reduction for comparison as well as IBAD method adopted in this study. Figure 4 shows surface roughness and morphology of surface electrode of IPMC prepared by those two methods. Figure 4 (a) shows the surface of IPMC plated with Pt by chemical reduction and Fig. 4(b) is the surface of IPMC plated by physical IBAD method.
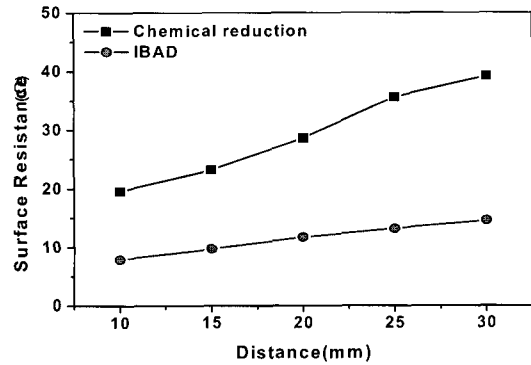

Fig. 5. Electric surface resistance of surface electrode of IPMCs with Pt plated by chemical reduction and by IBAD.

The surface roughness was ranged 0.4 [μm] by IBAD deposition, which indicates that the platinum layer became smoother and more homogeneous than conventional chemical reduction method. The effect of improved surface property could be reflected in electro-mechanical actuation performance of IPMC. Figure 5 shows the surface resistance of surface electrodes prepared by two processes. The surface electric resistance of chemically plated IPMC increases noticeably from 20 to 40 [Ω] along the IPMC strip, while the surface electrode by IBAD show as small as 8 [Ω] at end point and its slope along the strip is relatively quite low.

Actuation Characterization

IPMC with surface electrode made of precious metals such as Pt, Ir, and Au by IBAD were also prepared for comparison study. The electro-mechanical displacement and force of several IPMCs with different metal as surface electrode were measured. Cu and silver were also tried by IBAD deposition and was not proper as surface electrode due to their active reactivity in hydration condition. The thickness of Pt, Ir, and Au deposited on the surface of IPMC were approx. 500 [□]. Figure 6(a) and (b) shows the maximum displacement of the IPMCs and the initial actuation behaviors of the IPMCs according to induction time for

3 [min], respectively, when the voltage of 2 [V] DC was applied on the both faces of the IPMCs. The sample by chemical reduction (Pt) was also shown for comparison purpose. The displacement of IPMC plated with Pt by IBAD shows the greatest displacement among samples investigated. The Pt plating by IBAD endows the IPMC with rapidest reaction to displacement.
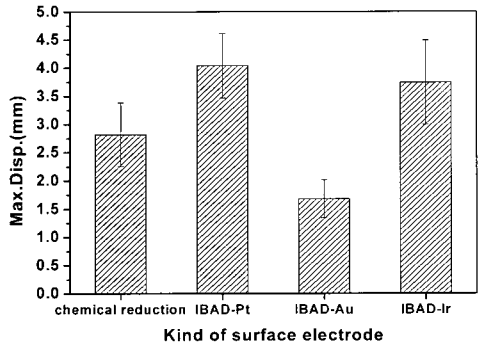
(a)
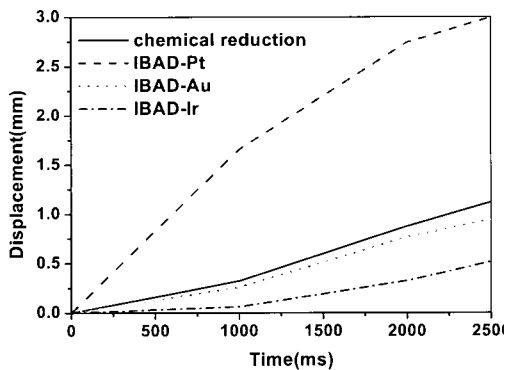
(b)

Fig. 6. Maximum displacement (a) and initial displacement (b) of IPMCs plated by chemical reaction (Pt) and IPMCs plated with different novel metals by IBAD for surface electroding.

Figure 7 shows the long time displacements of IPMCs plated with Pt through chemical reduction and by IBAD under electric voltage of 2 [V] DC in order to observe the back bending effect of IPMC. The mechanism of actuation IPMC is basically that hydrated cations move from anode toward cathode by electric stimuli. The composite in hydrated state performs an oscillatory bending motion when an alternating voltage is imposed across its faces. When constant voltage (DC) is applied across the faces of IPMC suddenly, an initial quick bending towards the anode is generally followed by a slow relaxation in the reverse direction towards the cathode [9-12]. The reasons of the relaxation can be electro-osmosis and transport of hydration cation. Chemically plated surface electrode shows the more relaxation in the

reverse direction, as shown in Fig. 7. On the other hand, a smoother and more homogeneous surface electrode imposed by IBAD appears to retard the relaxation definitely.
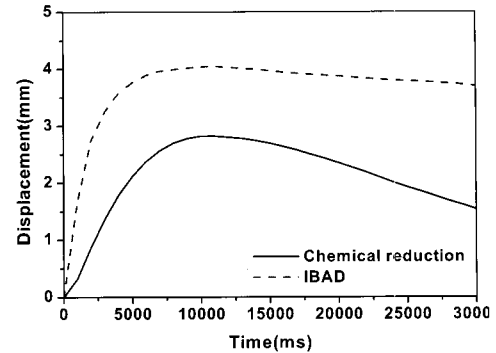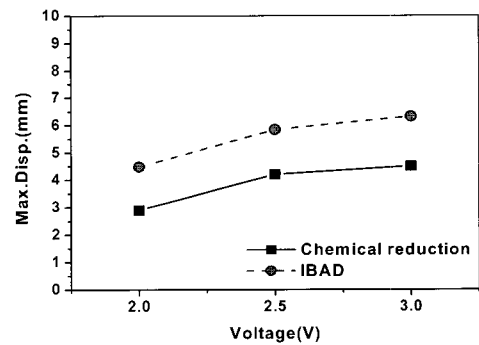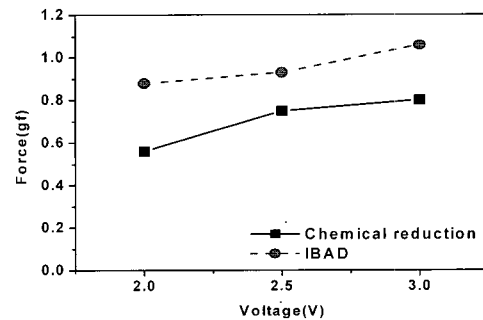

Fig. 7. Displacement of IPMC plated with Pt by chemical reduction (solid line) and by IBAD (dot line).

Figure 8 shows the actuation behaviors of IPMCs plated with Pt by chemical reduction and IBAD method, when the IPMCs were applied under voltage of 3 [V] AC at 0.1 [Hz] for 30 [sec]. The 0.1 [Hz] as frequency for AC voltage was chosen, because IPMCs showed the maximum displacement and force during the initial actuation for 30 [sec].

(a)

(b)

Fig. 8. Maximum displacement (a) and force (b) of the IPMC plated with Pt by chemical reduction and by IBAD.

The higher displacement and force of the IPMCs by IBAD is attributed to the effect of improved surface electroding. The maximum force of surface electrode by IBAD was over 1.0 [gf], while chemically plated composite was 0.8 [gf]. It was presumed partly due to the mechanical strength of composite increased by more homogeneous surface electrode, and the surface electrode density can be presumed as one of the factors for high actuating force of IPMC.

IPMC for Artificial Muscle

We used two types of Nafion membrane with different thickness for artificial muscle. Thin membrane of which thickness is about 178 [μm] was Nafion 117 (DuPont). Another thicker Nafion membrane was obtained by the solidification of Nafion solution (SE-10072, DuPont) at 80 [□] followed by the evaporation at 150 [□] to remove completely the Dimethyl formaldehyde (DMF) [13]. The thickness of the membrane was about 800 [μm]. For long-term actuation performance, each IPMC layer was coated and protected on the surface with selected chemical coat material.
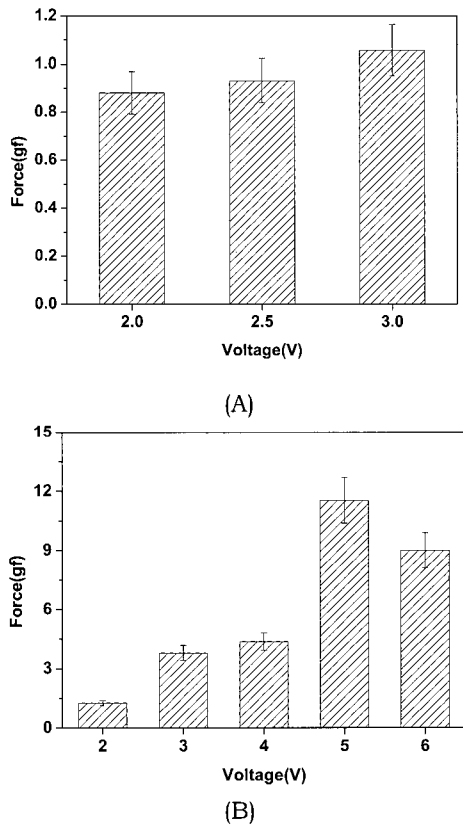
We prepared three types of IPMC actuators;
#1. IPMC specimen A (single layer, thickness: 178 [μm])
#2. IPMC specimen B (single layer, thickness: 800 [μm])
#3. IPMC specimen C (3 layer lamination, each layer's thickness: 178 [μm])

We first investigate the actuated bending force and displacement according to the input voltage (see Fig. 9 and Table 1). As a result, the bending force was up to the maximum 12[gf] from 1[gf] by actuating the single layered IPMC with 178 [μm], but the bending displacement was reduced to 6[mm] from 30[mm].

Table 1. Displacements of IPMC.

	period 1~2	period 3~4	average displacement
IPMC (A)	30 [mm]	31 [mm]	30.5 [mm]
IPMC (B)	6 [mm]	6 [mm]	6 [mm]
IPMC (C)	13 [mm]	11 [mm]	12 [mm]
IPMC (C*)	18 [mm]	20 [mm]	19 [mm]

(*IPMC (C) changed in sequence of layers)

Figures 10 to 13 show the actuation results of different IPMC samples. In each figure, the channel 1 indicates the electric current, and the channel 2 is the voltage. The channel 3 and 4 are control signals, respectively. Upper images show the configurations of IPMC actuators in each period. The x axis is time scale in which each period is 5 [sec]. The y axis of the 2, 3 and 4 channels are voltage scale (each period is 5 [V]), but that of 1 channel is current scale, (each period is 100 [mA], and it is 200 [mA] for Fig. 12.) Table 1 show the displacements of each period and average displacement, when each period is assigned 1 to 4 in sequence. When the power supply is shut from the IPMC actuators, the voltage imposed on the IPMC is found to decrease from 2 to 0 [V] gradually, since it carries the characteristic of a capacitor.
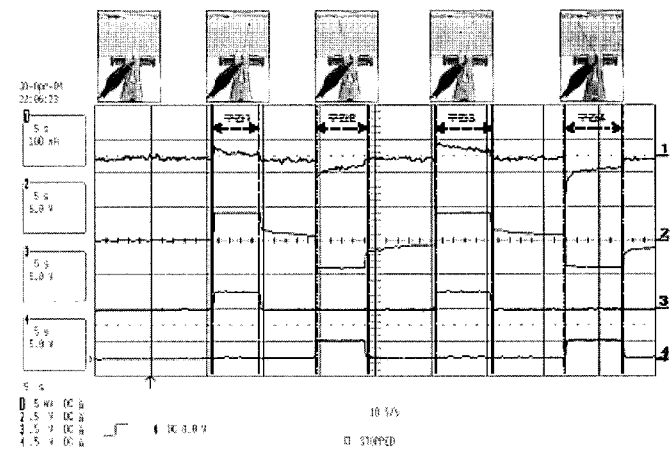

(A)

(B)

Fig. 9. Maximum forces of (a) a single IPMC with 178 [μm] thickness and (b) a single IPMC with 800[μm] thickness.

Fig. 10. Actuation of IPMC (A).

Fig. 11. Actuation of IPMC (B).

Fig. 12. Actuation of IPMC (C).
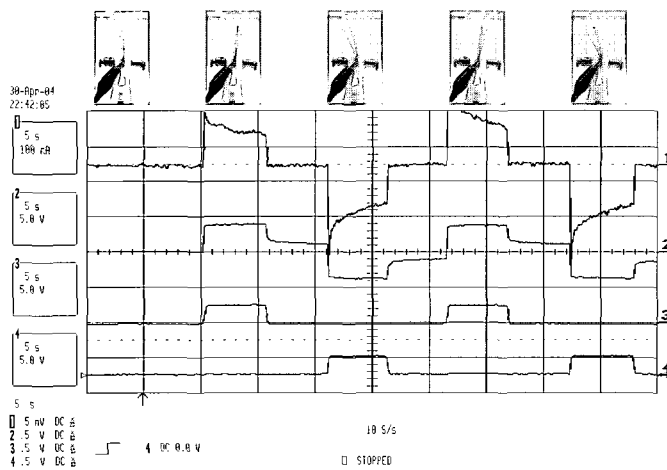
Fig. 13. Actuation of IPMC (C) changed in sequence of layers.

CONTROL OF IPMC BY EMGSIGNALS

In this study, we confirmed that the IPMC actuation can be controlled by human EMG signals, and it is considered as one of the essential features for biomimetic functions of IPMC for an application of artificial muscle. We first define the applied power polarities on both sides of IPMC to control the bending motions using EMG signals generated from the human flexor carpi ulnaris and extensor carpi ulnaris muscles. Muscle movements are recognized by comparing MAV signal with a threshold value. Figure 14 shows an example of motion classification method by the movements of flexor carpi ulnaris and extensor carpi ulnaris using a threshold. 'On' and 'off' mean whether the muscle movement is valid or not. Table 2 shows the muscle movement, corresponding states and applied power polarities. When extension motion occurs, the applied power polarities on side A is (-) and the applied power polarities on side B is (+). When flexion motion is occurred, the applied power polarities on side A is (+) and the applied power polarities on side B is (-). If no valid motion is occurred, the power is not applied.
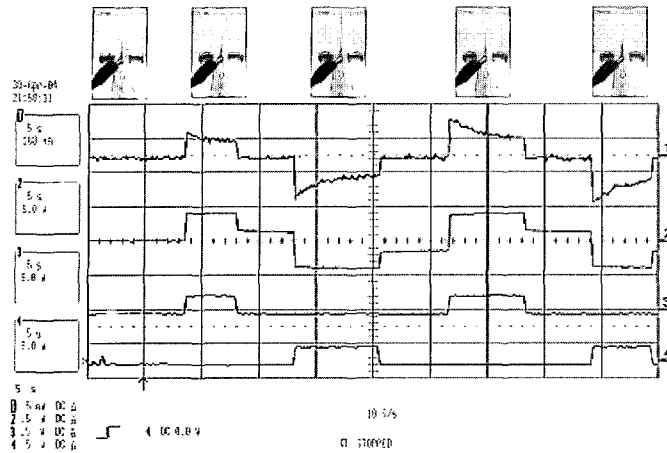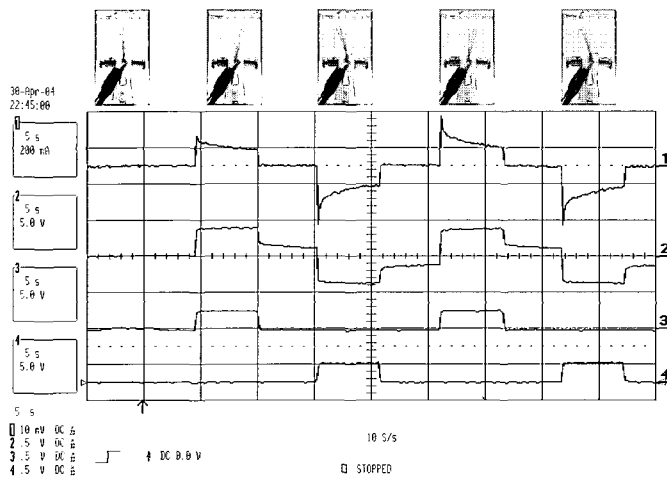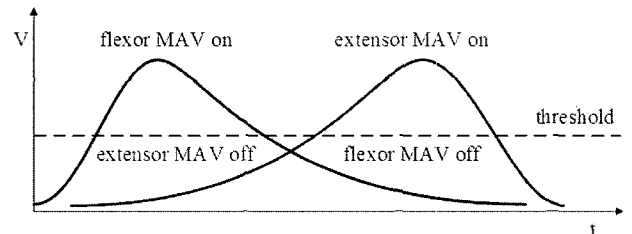

Fig. 14. Example of motion classification method by the movements of flexor carpi ulnaris and extensor carpi ulnaris using a threshold.

Table 2. Truth table of the power polarity on both sides of IPMC.

flexor MAV	extensor MAV	state	polarity	
			side A	side B
off	off	none	none	none
off	on	extension	(-)	(+)
on	off	flexion	(+)	(-)
on	on	none	none	none

Figure 15 shows the Actuation scheme of IPMC actuation controlled by EMG signals and its real image. Electric field and current imposed on the two faces of IPMC and two EMG signals entering the IPMC driver

circuit was measured with an oscilloscope (LeCroy 9354 TM). The measured data were transferred to PXI (PCI eXtensions for Instrumentation) using GPIB protocol. Real images of IPMC actuation captured with CCD camera were collected through PXI image collecting board (NI-1409).

The electric voltage imposed on IPMC was designed to change in a linear relation with the detected EMG signals, and thus the bending movement of IPMC is controlled linearly according to that of the wrist, as a real image of the actuation is shown in Fig. 16. Four strips shown in the figure were used to imitate the four fingers of human hands except thumb.
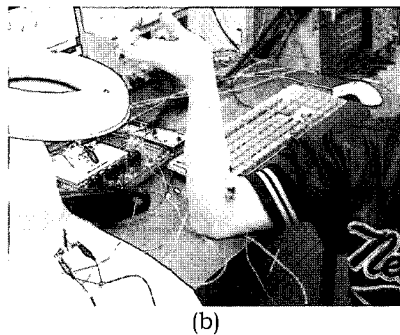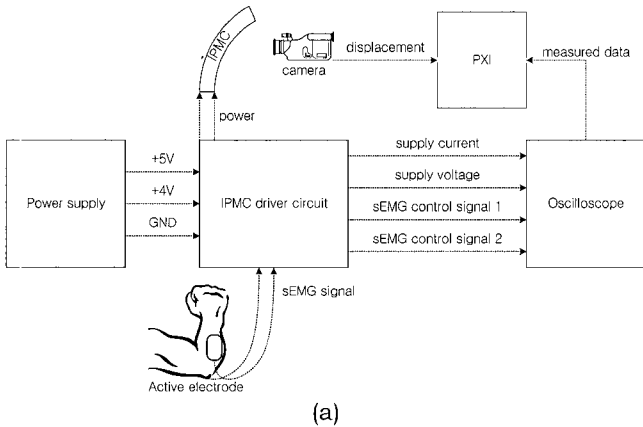
The electric field and current imposed on the two faces of IPMC and two EMG signals entering the IPMC driver circuit were measured with an oscilloscope for analysis. Figure 17 shows the control results by EMG signals generated from the human flexor carpi ulnaris and extensor carpi ulnaris muscles. The channel 3 and 4 are detected EMG signals, respectively. When the oversized voltage is applied on the IPMC, the IPMC driver circuit we designed limits the voltage imposed on the IPMC in order to prevent the abrupt physio-chemical change of the IMPC nature and subsequent shortening of the actuation operating time of the IPMC, as shown in the data for the channel 2 of Figure 17.

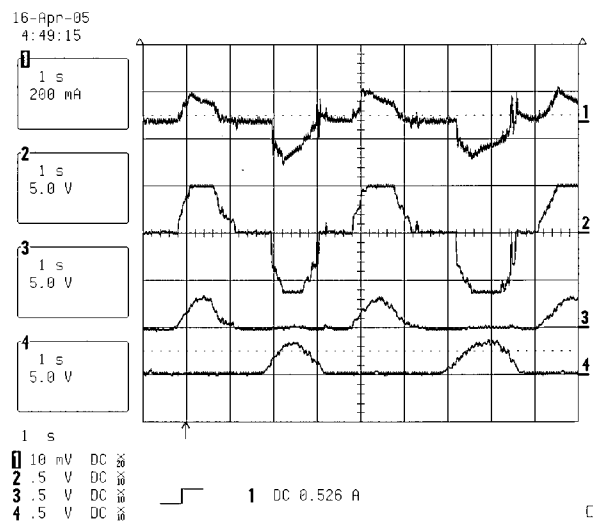**Fig. 15.** Actuation scheme of IPMC actuation controlled by EMG signals (a) and its real image (b).
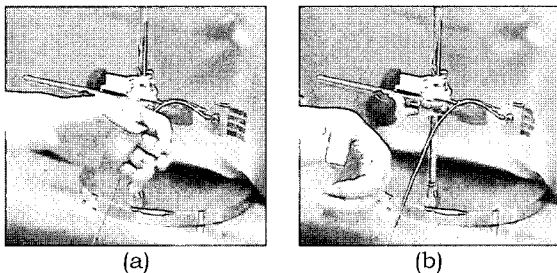
Fig. 16. Control of IPMC by EMG signal in none state (a) and flexion state (b).

Fig. 17. Control results by EMG signals generated from the human forearm muscles.

CONCLUSIONS

This paper proposed an IPMC control system with a bio-mimetic function. EMG signals generated by an intended contraction of muscles in forearm were used for the actuation of the IPMC. To obtain higher actuation force of the IPMC, the single layered as thick as 800 [μm] or multi-layered IPMC (Nafion) of which each layer can be as thick as 178 [μm] were prepared. To preserve the water existing within the IPMC for longer life performance, the each IPMC layer was coated and protected on the surface with some selected coating materials. To preserve the water existing within the IPMC for longer life performance, the each IPMC layer was coated and protected on the surface with some selected coating materials. It was successfully demonstrated that the IPMC actuation could be controlled efficiently with EMG signals generated from

the human flexor carpi ulnaris and extensor carpi ulnaris muscles. We also found out that the IPMC system can be implemented to confirm a possibility and a usability of the bio-mimetic artificial muscle controlled by human bio-signals.

In the future, we will focus on developing a biomimetic actuating system which can be used for the assistive devices by prosthetics and orthotics to assist the human muscle power in order to replace an electro-mechanical system like electric motors in the application field of rehabilitation technology and engineering.

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