A new approach to develop ionic polymer–metal composites (IPMC) actuator: Fabrication and control for active catheter systems

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Abstract

The ionic polymer–metal composite (IPMC) is one type of electro-active materials with the characteristics of low electric driving potential, large deformation and aquatic manipulation. It is highly attractive to biomedical applications as an actuator or a sensor. The main purpose of this study was to develop an IPMC actuator for active catheter systems.

The first step was to develop a low cost and high reliability fabrication procedure to yield an IPMC actuator. In the second step, the dynamic behavior of the actuator was tested in an aqueous environment. An empirical model was then constructed, which consisted of a fourth-order linear system, a nonlinear gain and a time delay. To linearize the dynamic behavior of this actuator for better actuating performance, a nonlinearity compensation method by a second-order polynomial was proposed. In the final step, the bending behavior of the constructed IPMC actuator with an open-loop and a closed-loop controller design was investigated.

The results indicated that a low cost but reliable IPMC actuator was fabricated successfully. Its production time was less than half of current manufacturing time (more than 48 h). The bending motion at low operation frequencies was well controlled by a conventional PID controller without adding complicated control algorithm. Our proposed algorithm decreased the maximum overshoot from 30 to 4.2%, and the steady-state error from 15 to 4%. Though the rise time was increased from 0.084 to 0.325 s, it was within the limit for many biomedical applications.

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1. Introduction

In recent years, cardiac catheterization has been becoming a common procedure for diagnosing or treating coronary heart diseases. To develop an active cardiac catheter, Guo et al. employed one type of electro-active polymers, ionic polymer–metal composite (IPMC), as the bending actuator with large deformation capability [1,2]. Because of its flexibility and low driving power, IPMC is highly attractive as a material for actuators or sensors in biomedical applications and for developing artificial muscles or biomimetic robots [3–5].

In general, an IPMC consists of a proton exchange membrane (PEM) with metal electrodes on both sides. It can undergo large bending deformation in response to an electric potential applied across its thickness in aqueous environment. The main bending mechanism is the unsymmetrical swelling of PEM caused by water migration with hydrophilic cation movement. Therefore, IPMC bends toward the anode of applied electric potential [2].

Current researches are focused on the actuation behavior of IPMC, such as electrochemical reactions and mechanical properties [5,6]. Equivalent circuit models of IPMC have been developed. The researches of displacement or force feedback control have been focused on how to handle the nonlinear and time-varying mechanical characteristics of an actuated IPMC [7–9]. The researchers focus on constructing an empirical model by using mechanical responses, and apply it on small vibration control in dry condition.

From the results of recent studies, design factors that can enhance the bending performance and reliability of IPMC are (1) low surface-electrode resistance, (2) high anti-oxidation,
compactness and flexibility of electrodes, (3) good adhesion of electrodes with the PEM, (4) high content of water or electrolyte in PEM, and (5) electrolyte with low vapor pressure [10–16]. To achieve the above-mentioned factors, some refined methods of fabrication have been proposed, such as plating platinum or gold electrodes on both sides of a PEM by employing impregnation-reduction technique (one kind of electroless plating art) [14], using Flemion® or other materials instead of Nafion® as PEM [15], and choosing a proper organic solvent as the electrolyte [16].

The common procedures of plating platinum/gold electrodes are to roughen the PEM surfaces, and then soak it in platinum/gold complex solution. In the next step, the PEM is immersed in a proper reducing solution to plate electrodes on its surfaces [13]. The foregoing method offers a good way to develop an IPMC actuator. However, the multi-plating requirements and long chemical reaction time of those procedures increases manufacturing cost. Since low cost is required for a disposable type of active cardiac catheter, as few as possible noble metals used for the electrodes and simplified fabrication procedures are mandatory.

In our previous work, an IPMC with silver-plated electrodes was developed, in which the adhesion of electrodes with PEM was enhanced by the pretreatment of dissolving and casting silver nano-powders in liquid PEM solution [17]. Moreover, a simplified stiffness model of IPMC was utilized to analyze the mechanical properties of the actuator. The advantages of our previous fabrication process were low cost and short processing time. However, the disadvantages were that oxidation of silver occurred easily during electrical actuation and degraded the bending performance, such as deformation and slew rate.

The goals of this work were three-fold. First is to design an IPMC actuator with a low cost and high reliability manufacturing process modified from our previous protocol [17,18]. Second is to verify an empirical model formulation of the actuator by a series of dynamic testing in an aqueous environment. In addition, a nonlinear compensation method is proposed to linearize the nonlinear dynamic behavior of the actuator. Final is to control the bending motion of the actuator by implementing a PID controller. The performance of open-loop and closed-loop control was compared and discussed.

2. Methods

2.1. IPMC fabrication principle and design

Electroless plating techniques of gold to form electrodes for IPMC are classified into three types by mechanisms, i.e. autocatalytic, galvanic displacement, and substrate-catalyzed processes [19–22]. The autocatalytic plating reaction can spontaneously deposit metal layers on any surface of a substrate within the bath. But the non-selective deposition and short lifetime causes considerable wasting. The galvanic displacement reaction occurs only on a metal substrate via the redox reaction of plated metal with the substrate. The plated metal requires higher reducing potential than the substrate. The limited deposition thickness and porous structure are the drawbacks. The substrate-catalyzed reaction is similar to the combination of autocatalytic and galvanic displacement reaction. It only deposits on a catalytic metal with low reducing potential but produces thicker compact deposition than galvanic displacement reaction. Because of the above reasons, a PEM with nickel layer is fabricated as the substrate for substrate-catalyzed reaction plating.

The main procedure of IPMC fabrication was carried out in three steps. The first was to form a composite of catalytic metal with PEM by mixing and baking nickel nano-powder with a liquid PEM, i.e. Nafion® solution (Table 1). In order to evenly disperse the nickel powder so that it could act as a good catalytic substrate, the cetyl trimethyl ammonium bromide (CTAB) was added. Besides, dimethyl formamide (DMF) was essential to enhance the mechanical property of this composite. Subsequently, this structure was immersed in a commercial electroless gold plating bath. Some well known substrate-catalyzed baths (e.g. hypophosphite and hydrazine baths [19]), could also be used. In the bath, a gold layer was deposited on the nickel side of the composite. The last step was to paint a thin layer of diluted PEM solution on polymer-side of the composite produced from the previous steps so that two pieces of the composite could be bonded together on the polymer-side by hot embossing.

The exact ingredients and steps of the procedure were summarized in the following (Fig. 1):

1. Mix 0.3 g of nickel nano-powder with 1 cm³ of 20% Nafion® solution in a glass dish of 8 cm in diameter, and bake at 50°C to form a nickel substrate. The additives in this process were 6 cm³ of CTAB (0.4 g/cm³) and 1 cm³ of DMF.
2. Pour 2 cm³ of 20% Nafion® solution onto the nickel substrate, and produce the nickel–Nafion® membrane by gradually baking from 50 to 120°C.
3. Cut the membrane into a piece of 6 cm × 1 cm and suspend it within a commercial gold plating bath (SHA-5, Sheng Hung, Taiwan) for 3 h to deposit a gold layer.
4. Split membrane into two pieces, and coat some diluted Nafion® solution (5–10%) on the polymer-side of both pieces. Then, perform hot embossing at a baking temperature of 110°C.

2.2. Experimental setup [18]

To investigate the actuating performance of an IPMC actuator, a testing system was established (Fig. 2). The main components are described in details in the following.
Fig. 1. Fabrication process of an IPMC structure.

Fig. 2. A sketch of experimental setup for bending performance testing.

A flexible clamp was used to hold the IPMC actuator and to transmit the electrical driving potential. A laser gauging sensor (Banner LG10A65PU, Minneapolis, MN, USA) was used to detect the tip displacement of the actuator without contact. The sensing range, precision, and bandwidth of the sensor were 0.2–50 mm, 200 μm and 450 Hz, respectively. The development system for controller (dSPACE DS1104, GmBH) sent commands through a power amplifier circuit to drive the actuator, and simultaneously received the displacement signal from the laser gauging sensor. The power amplifier circuit not only provided sufficient electric energy to drive the actuator, but also avoided the coupling between the actuator and the dSPACE control system.

In order to maintain persistent and steady drive, the IPMC actuator was suspended and submerged, except the clipped end, in a water tank filled with saline solution to sustain water uptake. All the experiments in this study were performed in this aqueous environment. To reduce the measuring error caused by the different refractive indices among air, glass tank and water, the tip displacement sensed by the laser gauging sensor was calibrated by using the precision multi-axis translation stage (Fig. 3). The

Fig. 3. The calibration result of the laser gauging sensor for aquatic manipulation.
calibration range was from 1 to 15 mm, and the measuring error was lower than 0.1 mm.

2.3. Definition of bending angle

Because of the sensing range limitation of the displacement sensor, large tip displacements could not be measured. To circumvent this problem, the displacement of a point (15 mm from the fixed end) along the IPMC shaft was measured and the displacement of the tip was estimated. It was assumed that the clipped end of IPMC actuator was perfectly fixed and the whole movable shaft of IPMC, under the actuated state, was an ideal arc (Fig. 4). A local bending angle, $\theta_C$, could be calculated from the geometric relationship between the sensing displacement, $X_d$, and the distance, $L_C$, measured from the clipped end to the detected point (Eq. (1)). The overall bending angle, $\theta_T$, was estimated by associating $\theta_C$ with the curvature radius, $R_C$ (Eqs. (2) and (3)):

$$\theta_C = 2\theta_L = 2 \tan^{-1} \left( \frac{X_d}{L_C} \right)$$  \hspace{1cm} (1)
$$R_C = \sqrt{\frac{(X_d)^2 + (L_C)^2}{2(1 - \sin(2\theta_L))}} = \frac{L_C}{\sin(2\theta_L)}$$  \hspace{1cm} (2)
$$\theta_T = \sin^{-1} \left( \frac{L_T}{R_C} \right) = \sin^{-1} \left( \frac{\sin(\theta_C) L_T}{L_C} \right)$$  \hspace{1cm} (3)

2.4. Experimental measurement, modeling, and control demonstration

To test actuation capability of the IPMC fabricated in this study, an IPMC actuator with a physical dimension of 20 mm in length, 5 mm in width and 250 $\mu$m in thickness was fabricated. The laser gauging system was positioned so that the distance from the clamped end to the detection point was fixed at 15 mm in all the following tests.

In the first test, the actuator was driven by a series of constant electric potentials, 0.5, 1.0, 1.5, 2.0, 2.5, and 3.0 V, respectively. The bending angle responses and currents were recorded for 30 s. At the end of each test, an intermission interval of about 10–30 min was allowed for recovering the actuator shape. In the second test, the driving potential was changed to sinusoid forms with similar amplitudes as those the first test. The adopted frequencies for sinusoidal driving were 0.1, 0.3, 0.5, 0.7, 1.0, 3.0, 5.0, 7.0, 10, 13, 15, 17, and 20 Hz. Then six sets of bending frequency responses were established in the form of Bode
diagrams. From the Bode diagrams, an empirical model was formulated, and a nonlinear compensation method was adopted to eliminate the gain nonlinearity in the system. They will be described in Section 3.

To overcome the relaxation and hysteresis problems of IPMC actuator, a closed-loop PID controller, \( G_{PID}(Z^{-1}) \), was designed (Eq. (4)), in which \( Z^{-1} \) was a operator of Z-transform, \( K_p \), \( K_I \), and \( K_D \) were controller parameters and \( T \) was sampling period. The specifications parameters of the controller performance were overshoot percentage and steady-state error less than 10 and 5\%, respectively. The reference trajectories were step, sinusoid, square and triangular waves for a bending angle of 5\(^\circ\) at 0.5 Hz:

\[
G_{PID}(z^{-1}) = K_P + K_I \frac{T(1 + Z^{-1})}{2(1 - Z^{-1})} + K_D \frac{1 - Z^{-1}}{T} \tag{4}
\]

3. Results

3.1. Fabrication results

Based on the comparison between failure and success fabrication results of nickel–Nafion® composites shown in Fig. 5, it was observed that the composite was less prone to crack with proper proportion of nickel powder to Nafion® solution and sufficient stir at 50\(^\circ\)C. In the next step, a compact and uniform nickel–Nafion® composite was constructed as the gold plating substrate. After the plating process, a sheet of half IPMC with dark gold electrode was formed (Fig. 6). The approximate sheet resistance value, \( R_S \), of electrode was 2–4 \( \Omega \) cm/cm as shown by Eq. (5), where \( \sigma_r \) was the resistivity, \( R \) was the resistance, \( L \), \( w \) and \( t \) were length, width and thickness of the electrode layer,

\[
R_S = \frac{\sigma_r}{t} = \frac{Rw}{L} \tag{5}
\]

From the cross-sectioned scanning electron microscopy (SEM) micrograph of an IPMC sample (Fig. 7), it was clear that gold was mainly deposited on the outside surface of the nickel layer. The overall thickness was about 250 \( \mu \)m, in which, the nickel and gold layers were about 15 and 5–20 \( \mu \)m, respectively. The IPMC actuator produced by our proposed fabrication process behaved as expected. Its performance was demonstrated by applying electric potential of 3 V at 0.1 Hz (Fig. 8).

3.2. Experimental test results

From the step responses of the IPMC actuator (Fig. 9), the maximal current induced by different driving potentials was less than 0.08 A. The induced current increased non-proportionately with the driving potential and the increase was larger at larger electric potentials, e.g. 2.5 and 3.0 V. For the bending angle, the bending magnitude increased with the amplitude of driving potential from 0.5 to 3.0 V. The maximal bending angle was about 20\(^\circ\). The bending angle decreased slowly over time during
Fig. 9. Step responses of the IPMC actuator with constant driving potentials. The upper diagram shows the amplitudes of driving potentials, the middle shows the induced electric currents, and the bottom shows the bending angles in response to the driving potentials.

The phase of constant driving potential and the decrease was more evident in higher driving potentials.

The frequency responses of bending angle, in the form of Bode diagrams, are shown in Fig. 10. All the curves showed two resonant peaks at 3–4 and 18–20 Hz, and their phase plots were almost identical, and the maximal phase was about 500°. Moreover, the plot also revealed that the bending angles in response to unit driving potential at different driving amplitudes were different. To eliminate this inequality, a curve was constructed according to the relationship between the averaged maximal bending angles and driving potentials in the range of 0.1–1.0 Hz (Fig. 11). The curve was fitted with a second-order polynomial.

3.3. Empirical model and bending angle control

From the experimental results mentioned above (Figs. 9 and 10), the empirical model consisted of a transfer function \( T(s) \), a time-delay term \( e^{-\tau f} \), and a nonlinear gain \( H_V\theta(V(t)) \), in which \( \tau \) was time delay and \( V(t) \) was driving potential. \( T(s) \) was unknown but modeled as a linear system with fourth or higher order on the basis of two resonant peaks shown in Fig. 10. The time-delay term, \( e^{-\tau f} \), was neglected because of low driving frequency in our applications, and \( H_V\theta(V(t)) \) was estimated from the curve in Fig. 11 (Eq. (6)):

\[
H_V\theta(V(t)) \approx 0.8345V^2 + 3.690V
\]  

(6)

The feedback control design was shown in the block diagram (Fig. 12), in which \( \theta_r(t) \) and \( \theta_p(t) \) were reference input and measured bending angle of IPMC actuator. And the inverse of \( H_V\theta(V(t)) \) was estimated from the curve in Fig. 11 (Eq. (7)). It was a compensation term utilized to neutralize \( H_V\theta(V(t)) \):

\[
H_V^{-1}(\theta(t)) \approx -0.0029\theta^2 + 0.2122\theta
\]  

(7)

The results of open-loop and closed-loop control actuated with step reference input 5° were shown in Fig. 13. The open-loop controller with the nonlinearity compensation term showed that the maximum overshoot was about 30%, the steady-state error was about 15%, and the rise time was about 0.084 s. On the other hand, the PID closed-loop controller (\( K_P = 0.4, K_I = 0.8 \) and \( K_D = 0.08 \)) decreased the maximum overshoot percentage and the steady-state error to 4.2 and 4%, respectively. Yet, the rise time was prolonged to 0.325 s. The low frequency responses of the open-loop controller are shown in Fig. 14. The tracking error was smaller for the sinusoidal reference input. On the contrary, the bending angle responses in the closed-loop controller conformed to the reference input trajectories well (Fig. 15). In general, the errors of closed-loop controller were smaller.
4. Discussion

4.1. Fabrication

From the fabrication results shown in Fig. 5, internal stress was likely the main factor causing cracks. A Nafion® membrane has a large strain property along with abundant water content. Hence, an improper compound proportion of nickel to Nafion® might cause the composite to crack with uneven internal stress distribution. With more powder and less Nafion® solution, the composite might crack due to insufficient stickiness. With less powder and more Nafion® solution, the nickel layer was prone to have many pinholes that would decrease the electric conductivity of gold layer to be plated in subsequent steps. These were the reasons to separate the fabrication procedure of nickel-IPMC substrate into first step of more powder but less Nafion® solution and second step of less powder but more Nafion® solution. After the gold plating step, the surface of gold electrode was covered with a thin residual Nafion® polymer (Fig. 6). The thin Nafion® layer could be easily scraped to expose the underneath gold surface for conducting electric driving potential.

The production time of the fabrication procedure proposed in this work was less than 24 h. It was just 50% of the time for the existing manufacturing procedures in the literature (more than 48 h) [2]. The possible causes of shorter production times included: (1) the nickel layers on IPMC effectively decreased the water effusion, so that thick gold electrodes was not required; (2) in general, the substrate-catalyzed plating was faster than impregnation-reduction plating. The cost of gold plating in this work was also less than most of the existing manufacturing procedures without sacrificing the plating quality.

4.2. Testing, modeling, and control

Based on the experimental results shown in Fig. 9 and data of other literatures [6–9], IPMC actuators were nonlinear and time-varying systems. Yet, the aqueous manipulation alleviated the response relaxation problem. The reasons might be the distribution of electric field lines, and the gradient variation of cation concentration. The intensity of electric field was increased along with raising the driving potential, and more cations with water accumulated at one side of IPMC and cause the actuator to deform. However, the deformation might also change the distribution of electric field, and the cations with water might effuse out from the surface pores of the expanding side. Because of this interaction, the cation distribution in IPMC associated with an electric field was a very complex nonlinear phenomenon, and the effusion of cations with water resulted in the relaxation (drift with time) phenomenon.

According to the Bode diagram (Fig. 10), the second peak resonant frequency was ~5–6 times of the first peak resonant frequency, and the slopes of the magnitude plot between two
peaks were about $-40$ dB/decade. It implied that the actuator dynamic behavior could be approximated by a second-order system, even though the damping ratios were different in different driving potentials. Therefore, it was expected to have a good performance by implementing a PID feedback controller. However, phase-lag/time delay, as shown in the phase plot, might cause the closed-loop control to become unstable at higher actuating frequencies. The feasible actuating frequency range was initially set at less than 1 Hz.

From the averaged maximal bending angle exhibited in Fig. 11, the nonlinear gain of the empirical model could be estimated and compensated by inverting the estimated nonlinear gain curve. The nonlinear behavior of the actuator was mostly compensated, as shown by the results of open-loop control with constant and low frequency command (Figs. 13(top) and 14). The bending angles approached the reference trajectories closely. Yet, closed-loop control was still necessary to overcome the bending angle drift problem and to regain zero state position.

In contrary to the empirical models that other studies constructed in the air state and might behave different from the actual operation condition, the model in this study was constructed for aqueous manipulation. So the states of cation concentration and water content in actuator were controlled. Therefore, the reliability of the model formulation was higher than in dry environment. After compensating the nonlinear effect by a second-order polynomial approximation, a PID controller effectively achieved the control goal. Using the nonlinearity compensation could reduce the cost of controller design. Despite the time-delay problem was not treated, the IPMC actuator could work in low frequency range as shown by the control results (Figs. 13 and 15).

For the development of an active catheter system, shape memory alloy (SMA) and magnetic types of actuators have been proposed but were found to have some drawbacks, e.g. highly temperature dependency, requirements of sophisticated manufacturing and driving equipments. On the contrary, the IPMC actuator has the potentials of large bending range, high controllability, low heat generation, and easy manufacturing. Therefore, it has a great potential to be implemented as an active catheter in the future.

In spite that both the model and the controller were simple and conventional, the test results indicated that the performance of the model and the controller was good. Actually, we think one of our main contributions was to be able to reduce the complex system into a simpler one so that it could be controlled by a simple controller. In the future, for applications demanding a higher frequency band of operation, we plan to construct a more precise dynamic model and employ a robust controller for the bending motion control.

5. Conclusion

A low cost and reliable IPMC actuator applicable to the development of disposable active cardiac catheter was designed and fabricated successfully. A proper proportion of nickel powder with Nafion® solution was required to synthesize a nickel–Nafion® substrate. An empirical model of the IPMC actuator for aqueous manipulation was constructed. The model consisted of a fourth-order transfer function, a nonlinear gain and a time-delay term. Its parameters were varied with driving potential, operating time and frequency. With the nonlinear compensation method proposed in this work, the model was linearized and the actuator performed well under the conventional PID closed-loop controller in bending angle control experiments.

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References

Biographies

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