Modeling and Experiments of IPMC Actuators for the Position Precision of Underwater Legged Microrobots

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Abstract — Nowadays, underwater microrobots show significant potential of monitoring ocean currents and chemical agents, studying the animal migration, depth measurements, pollution detection, and video mapping in limited space. Compact structure, multi-functionality, flexibility, and position precision are normally considered incompatible characteristics for underwater microrobots. To accomplish our objectives, we designed several bio-inspired underwater microrobots with compact structure, flexibility, and multi-functionality by using ionic polymer metal composite (IPMC) actuators. However, the position precision of them was not high enough. To implement a high position precision of developed underwater microrobots, we proposed an electromechanical model of the IPMC and analyzed the deformation and actuating force of the IPMC equivalent cantilever beam, which could be used as biomimetic legs, fingers or fins of the underwater microrobot. Then we evaluated the tip displacement of the IPMC actuator experimentally. The experimental deflections fitted the theoretical values very well when the driving frequency was larger than 1 Hz. In addition, by using the IPMC equivalent cantilever beam model, we developed several underwater legged microrobots. A predetermined trajectory tracking experiment was carried out to evaluate the position precision of developed legged microrobot.

Index Terms – IPMC actuators, Electromechanical model, Position precision, Underwater biomimetic microrobot.

I. INTRODUCTION

The underwater microrobots can be widely used in underwater monitoring operations, including pollution detection, video mapping, and exploration of unstructured underwater environments. However, the electromagnetic structure of traditional motors is difficult to miniaturize. Thus, motors are rarely found in this type of application [1–2], and special actuator materials are used instead. A variety of smart materials, such as ionic polymer metal composite (IPMC), piezoelectric elements, pneumatic actuators, and shape memory alloy, have been investigated for use as artificial muscles in new types of microrobot [3–4].

The actuation characteristics of the IPMC, including a suitable response time, high bending deformation, and long life, show significant potential for the propulsion of underwater microrobots, which can be used as artificial muscles to propel robots backwards and forwards. [5]. So, they are widely used as oscillating or undulating fins for swimming microrobots when fast response is required [6]. The typical representative researches focused on the fundamental properties and characteristics, manufacturing techniques, phenomenological modelling of actuators, and sensing mechanism [7]. Yun and Kim proposed three fingers gripper in which each finger was actuator was actuated individually [8]. Bonomo et al. offered a nonlinear dynamic model based on a gray box [9]. Gong et al. developed a Finite Element (FE) model for simulating the dynamic electromechanical response of an IPMC structure under an external voltage input [10]. Although a large amount of research activity has been devoted to the study of IPMC–based actuator, the deformation and generated force are still under investigation and a general consensus does not exist [9].

We also developed several underwater legged microrobots with efficient locomotion and multi-functionality, which employed IPMC-based biomimetic locomotions to implement underwater walking, rotating, floating, and swimming motions [11–15]. However, the position precision of IPMC actuators were not high enough, which was essential in some simple tasks such as detecting an object, grasping and carrying objects to a desired position, or avoiding an obstacle. To implement a high position precision of developed underwater microrobots, we proposed an electromechanical model of the IPMC and analyzed the deformation and actuating force of the IPMC equivalent cantilever beam, which could be used as legs, fingers or fins of the microrobots.

The remainder of this paper is divided into five parts. First, we describe the definition of an electromechanical model, including the electrical part, theoretical deflection characteristics and theoretical force characteristics. Second, we measure the deflection of the IPMC sample experimentally to evaluate the proposed theoretical model. Third, based on above electromechanical model, we present the design of a conventional PID controller to implement high position precision of IPMC actuators. Fourth, we introduce the predetermined trajectory tracking experiment of a previous developed legged microrobot to evaluate its walking and rotating precision. Finally, we present our conclusions.

II. THE ELECTROMECHANICAL MODEL

The IPMC actuators can be treated as equivalent cantilever beam. Figure 1 shows the mechanical configuration of the IPMC actuator and relevant parameters. Then we show the configuration and the relevant geometrical parameters in
the following: \( L_c \) denotes the length of the clamped part of the IPMC; \( L_f \) is the total free length of the IPMC; \( w \) and \( h \) denote the width and the height of the IPMC cross section. The pinned end is used to apply the electrical voltages across their thickness direction.

According to mechanical analysis, the bending deformation of IPMC actuator is resulted from internal water molecules redistribution. Under the influence of applied stimulus, the water molecules in the IPMC actuator are redistributed by following two stages [10]:

(1) When the electrical stimulus is applied across IPMC’s thickness direction, each hydrated sodium ion moves together with four hydrated water molecules to the cathode side. Then the bending deformation is generated for the swelling of Nafion117 near the cathode side and contraction near the anode side.

(2) In a short time, due to the self-diffusion of water molecules, the free water molecules flow to the anode side gradually, which reduces the concentration of water molecules at the cathode and shows a deformation recovery potential of the IPMC actuator.

The model of IPMC-based actuators is divided into two stages. The external stimulus to the model is the imposed voltage \( V(t) \), while the first stage output is an estimation of the absorbed current \( I_1(t) \). As widely reported in the literature, the current produces the IPMC mechanical reaction because of charge/water redistribution [9]. The second stage is intended to estimate either the available force \( F(t) \) or the tip displacement \( \delta(t) \), in the absence of an external force.

A. The Electrical Part

For the IPMC actuator is driven by the electrical voltage, so it shows some electrical characteristics. Then we use an equivalent RC circuit to model the IPMC actuator, which is able to convert the applied voltage stimulus into the inner current. This RC model is used for determining the electric charge produced by an input voltage. This inner current is in fact a redistribution of inner ions which generates an electrical field across the thickness direction of IPMC actuator. The equivalent RC circuit model with lumped parameters shows some advantages for it can provide a graphical representation of the governing equations for an IPMC leg or finger in real applications. The electrical elements used in the RC circuit can be evaluated based on physical considerations which allow scaling them by the IPMC geometry [9].

Figure 2 shows the equivalent lumped RC circuit which is adopted for the IPMC electrical model. In this equivalent electrical circuit, \( R_c \) denotes the resistances of two electrodes, \( R_1 \) denotes the equivalent bulk resistance of the Nafion 117, and \( R_C \) reflects the capacitive nature of the IPMC.

According to the Kirchoff’s voltage law, we can get the equations (1).

\[
V(t) = R_1[I(t) - \frac{V(t) - 2R_1I(t)}{R_1}] + 2R_1I(t) \\
+ \frac{1}{C} \int [I(t) - \frac{V(t) - 2R_1I(t)}{R_1}] dt
\]

where \( V(t) \) denotes the external stimulus, \( I(t) \) denotes the absorbed total current, \( I_1(t) \) denotes the current across \( R_1 \), \( I_2(t) \) denotes the current across \( R_C \).

It is assumed that there is no initial current flow. Then we take the Laplace transformation for equation (1) and get equations (2) and (3).

\[
V(s) = R_1[I(s) - \frac{V(s) - 2R_1I(s)}{R_1}] + 2R_1I(s) \\
+ \frac{1}{sC} [I(s) - \frac{V(s) - 2R_1I(s)}{R_1}]
\]

\[
I_1(s) = \frac{s(R_1C + R_C) + 1}{s(R_1R_C + 2R_1R_C + 2R_1R_C + R_1 + 2R_C)}V(s)
\]

The basic charge equation is shown as (4).

\[
Q(t) = \int I_1(t) dt
\]

It is assumed that there is no initial current flow. Then we take the Laplace transformation for equation (4) and get equation (5).

\[
Q(s) = \frac{I_1(s)}{s} = \frac{s(R_1C + R_C) + 1}{s^2(R_1R_C + 2R_1R_C + 2R_1R_C + R_1 + 2R_C)}V(s)
\]

B. Theoretical Deflection Characteristics

The current absorbed by the IPMC is the cause of the mechanical reaction because of inner charge/water molecule redistribution, which generates a mechanical bending of the IPMC. The dynamic bending displacement of an IPMC beam \( \delta(t) \) is determined by the concentration of water molecules \( W(t) \), as shown in equation (6).

\[
\delta(t) = k_W (W(t) - 4k_Q(t))
\]

where \( k_W \) is the deformation coefficient of IPMC and \( Q(t) \) denotes the total electric charge. Under the saturated state, each sodium ion combines with four water molecules to form...
a hydrated sodium cation. So, the \( W(t) \) can be got as \( 4Q(t) \). Also, we assume that there is no initial current flow and deformation. Then we take the Laplace transformation for equation (6) and get equation (7).

\[
\delta(s) = sW(s) = 4kQ(s) \tag{7}
\]

Substituting \( Q(s) \) from (7), we can get equation (8).

\[
\delta(s) = 4k_s\frac{s(R_c + R_e)C + 1}{s(R_c + R_e)C + (R_e + 2R_c)}V(s) \tag{8}
\]

Then we will scale the elements in the IPMC equivalent circuit and the geometrical dimensions are shown as follows: \( l_f=17 \text{ mm}, \ l_c=3 \text{ mm}, \ w=4 \text{ mm}, \ h=0.22 \text{ mm}. \) \( R_e \) is a resistance of the electrode. With the same thickness and area of two electrodes, \( R_e \) is the same for both the electrodes of an IPMC actuator. By modeling each electrode as one layer with same thickness, we assume that its resistance is proportional to the free length of the IPMC and inversely related to its width. \( R_c \) can be determined by equation (9) [9].

\[
R_c = \frac{R_L l_f}{w} \tag{9}
\]

where \( R_c \) is induced resistance which can be estimated by performing adequate measuring surveys and processing data. Nation® \( \text{Na}^+ \) is used in our research, and the geometrical dimensions of IPMC are shown as follows: \( l_f=17 \text{ mm}, \ l_c=3 \text{ mm}, \ w=4 \text{ mm}, \ h=0.22 \text{ mm}. \) So, \( R_e \) can be approximately chosen as 1.075 \( \Omega \) [9]. Then \( R_c \) can be calculated as 4.6 \( \Omega \). From equation (9) we can see that the ratio \( R_c/R_e \) depends only on the sample geometrical dimensions.

\( R_f \) is the equivalent bulk resistance of the Nafion® in DC conditions. It can be computed as equation (10) [9].

\[
R_f = \frac{\rho_L l_f}{(L_f + L_e)w} \tag{10}
\]

where \( \rho_L \) denotes the Nafion® DC resistivity and \( h, \ L_f, \ L_e, \) and \( w \) are the geometrical dimensions shown in Fig. 1. \( R_f \) can be found on the basis of experimental data. Here with same IPMC sample, \( R_f=182.1 \text{ K}\Omega \).

\( R_2 \) denotes the equivalent bulk resistance of the Nafion® against the charges involved in fast phenomena. It was modeled as a function of both the Nafion® resistivity \( \rho_2 \) and the geometrical dimensions of the sample, as shown in equation (11) [9]. \( R_2 \) can also be found on the basis of experimental data. Here with same IPMC sample, \( R_2=0.6523 \text{ K}\Omega \).

\[
R_2 = \frac{\rho_1 h}{(L_f + L_e)w} \tag{11}
\]

The capacitor of the same branch \( C \) is scaled as equation (12). The value of the permittivity \( \varepsilon_2 \) can be found on the experiments [9]. For the same IPMC sample, \( C=0.04518 \text{ F/s}. \)

\[
C = \varepsilon_2(L_f + L_e)w \tag{12}
\]

The deformation coefficient \( k_e \) is tested with a value approximately equaling to 0.06875 for the IPMC sample. We assume the external stimulus \( V(t)=4(t) \), and we can get \( V(s)=4/s \). Taking inverse Laplace for equation (8), we can get equation (13), where \( \alpha=0.0334598685 \), \( \alpha_0=0.000121113 \). The tip deflection of the IPMC sample with time is shown in Fig. 3.

\[
\delta(t) = 0.00166876 \left( \frac{a_0 - \alpha}{\alpha^2} e^{-\alpha t} + \frac{a_0}{\alpha} + \frac{a - a_0}{\alpha^2} \right) \tag{13}
\]

\[= 0.0497e^{-0.0334598685t} + 6.0403 \times 10^{-4} t - 0.0497\]

![Fig. 3 Theoretical deflection of IPMC with time (step stimulus)](image)

**C. Theoretical Force Characteristics**

Figure 4 shows the electro-mechanical behavior of a cantilevered IPMC actuator under the electric field which was modeled as a supported cantilever beam under a uniformly distributed bending moment. According to the tip deflection equation (14) under distributed bending moment, we can obtain the equivalent resultant moment at the tip point, as shown in equation (15).

\[
\delta_x(t) = \frac{M_x(t) \cdot x}{EI} \left( l_f - \frac{x}{2} \right) \tag{14}
\]

\[
M_x(t) = \int M_x(t)dx = F_x(t)l_f \tag{15}
\]

Substituting \( M_x(t) \) from (15), we can get equation (16).

\[
\delta_x(t) = \int \frac{M_x(t) \cdot l_f^3}{2EI} \text{d}t = \frac{F_x(t) \cdot l_f^3}{3EI} \tag{16}
\]

We take the Laplace transformation for equation (16) and get equations (17). From (17), the resultant bending moment and equivalent force at the tip point can be calculated by equations (18) and (19).

\[
\delta(s) = \frac{M(s) \cdot l_f^3}{EI} = \frac{F_x(s) \cdot l_f^3}{3EI} \tag{17}
\]

\[
M(s) = \frac{s \cdot 2EI}{l_f^3} \delta(s) \tag{18}
\]

\[
F_x(s) = \frac{3EI}{l_f^3} \delta(s) \tag{19}
\]

We assume the external stimulus \( V(t)=1(t) \), and we can get \( V(s)=1/s \). The geometrical dimensions of IPMC sample
are as follows: \( l = 17 \) mm, \( w = 4 \) mm, \( h = 0.22 \) mm. The measured value of the elastic modulus \( E \) for IPMC under hydrated conditions is about 83 MPa [16]. For cross-sectional dimensions of 0.22 \( \times 4 \) mm, the moment of inertia \( I \) of IPMC is \( I = \frac{wh^3}{12} = 3.574 \times 10^{-15} \) m\(^4\). Taking inverse Laplace for equation (19), we can get equation (20).

\[
F_c(t) = 0.18114 \times 0.00166876 \left( \frac{\alpha_1 a^2 - \alpha_2 a}{a^2} e^{-\frac{a}{a^2}} + \frac{\alpha_2}{a} + \frac{a - \alpha_2}{a^2} \right) \tag{20}
\]

where \( a \) is the radius of the IPMC actuator, \( \alpha_1 \) and \( \alpha_2 \) are constants, and \( F_c(t) \) is the driving force at time \( t \).

III. EXPERIMENTAL DISPLACEMENT OF THE IPMC ACTUATOR

To evaluate the proposed electromechanical model of IPMC actuator, we measured the displacement of a single IPMC actuator by applying different signals in a water tank. Figure 5 shows the displacement measuring system. The actuator was driven by a PC equipped with a digital-to-analogue converter card, and the deflection of the IPMC was measured by a laser displacement sensor. The laser sensor was used to translate the displacement to a voltage, and then the voltages were recorded and translated to the PC using an oscilloscope. The sample of IPMC actuator was 20 mm long, 4 mm wide, and 0.22 mm thick.

IV. POSITION CONTROLLER DESIGN

A conventional PID controller was made to verify the tracking performance and robustness. The conventional PID controller, which was used to control the IPMC actuator, was built by the Simulink of MATLAB. With the PID controller, the control signal for the IPMC actuator can be expressed in the time domains as equation (21).

\[
v(t) = k_p e(t) + k_i \int_0^t e(t) dt + k_d \frac{de(t)}{dt} \tag{21}
\]

where \( e(t) \) is the error between desired reference signal and the output, \( v(t) \) is the control signal used to adjust the tip position of the IPMC; \( K_p, K_i \), and \( K_d \) are the proportional gain, integral gain and derivative gain of the PID controller, respectively. The PID gains in (21) were tuned by trials and errors to get acceptable control performance. Consequently, the most suitable PID parameters obtained by this method were \( K_p = 5891117, K_i = 27585109, \) and \( K_d = 147088 \). A step reference signal and a sine reference signal were applied to investigate step and sine responses of the IPMC system. The PID control results were displayed in Fig. 8 and Fig. 9 as the blue lines.

V. BIOMIMETIC MICROROBOTS

By using the proposed equivalent cantilever beam model we developed several underwater microrobot with compact structure and multi-functionality. Figure 10 shows the eight...
legged walking prototype. It is 33 mm long, 56 mm wide, and 9 mm high. It has eight IPMC actuators designated A through H, which are fixed on a film body with wood clips. Actuators A, B, C, and D are the drivers. The other four actuators are supporters. The IPMC actuators are all 11 mm long, 3 mm wide, and 0.2 mm thick. The distance between two adjacent drivers or between a driver and a supporter is 10 mm.

To evaluate walking and rotating locomotion, we carried out an experiment on an underwater plastic surface. When walking or rotating, the phase of the supporters lagged that of four drivers by 90°. Drivers A and B were driven by the same square wave, while C and D were also driven by another same square wave. The walking and rotating speeds were determined by the tip displacements of four drivers and the frequency of the square waves. With different deflections of four drivers between its two sides, it could rotate with different radii. It showed significant potential of tracking the predetermined trajectory to arrive at desired place. Figure 11 shows the tracking experiment on the underwater flat. The displacement of the IPMC actuator would be less in real applications due to loading and slippage. Therefore, some differences between the predetermined trajectory and experimental results existed as shown in Figure 12.

\[
\begin{align*}
\Delta d_1 &= \theta^* R_1 = \theta^* R_1 \left[ 28 - \left( 1 - \frac{d_1}{2} \right)^2 \right]^{1/2} \\
\Delta d_2 &= \theta^* R_2 = \theta^* R_2 \left[ 28 - \left( 1 - \frac{d_2}{2} \right)^2 \right]^{1/2}
\end{align*}
\]

where \( \Delta d_1 \) and \( \Delta d_2 \) are the tip displacements without payloads of four drivers between the microrobot’s two sides, \( \Delta d_1 \) and \( \Delta d_2 \) stands for the displacement decrease for the payloads in real application, \( l \) is the length of IPMC driver, and \( r_1, r_2 \) are the bending radii of the IPMC drivers.
VI. CONCLUSIONS

In this paper, we proposed an electromechanical model and analyzed the deformation and actuating force of the IPMC equivalent cantilever beam to ensure the position precision of IPMC actuator, which could be used for the biomimetic locomotion. Then the deflection of IPMC actuator was measured experimentally to evaluate the theoretical model. From the comparison, the experimental deflections fitted the theoretical values very well when the driving frequency was larger than 1 Hz, which showed the availability of the IPMC model to be used as oscillating legs or fins of the underwater microrobots. In addition, a conventional PID controller for IPMC actuators was designed from the identified model. Based on the IPMC electromechanical model, we carried out the trajectory tracking experiment by using previous developed legged walking microrobot on an underwater plastic surface. According to the predetermined trajectory, we calculated the desired deflection of four drivers and then obtained the driving voltages for each step. The microrobot was driven with an open-loop control and some differences between the predetermined trajectory and experimental results existed for the slippage of four drives.

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