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REVIEW ARTICLE

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REVIEW ARTICLE

Ionic polymer–metal composites (IPMCs) as biomimetic sensors, actuators and artificial muscles—a review

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Abstract. This paper presents an introduction to ionic polymer–metal composites and some mathematical modeling pertaining to them. It further discusses a number of recent findings in connection with ion-exchange polymer–metal composites (IPMCs) as biomimetic sensors and actuators. Strips of these composites can undergo large bending and flapping displacement if an electric field is imposed across their thickness. Thus, in this sense they are large motion actuators. Conversely by bending the composite strip, either quasi-statically or dynamically, a voltage is produced across the thickness of the strip. Thus, they are also large motion sensors. The output voltage can be calibrated for a standard size sensor and correlated to the applied loads or stresses. They can be manufactured and cut in any size and shape. In this paper first the sensing capability of these materials is reported. The preliminary results show the existence of a linear relationship between the output voltage and the imposed displacement for almost all cases. Furthermore, the ability of these IPMCs as large motion actuators and robotic manipulators is presented. Several muscle configurations are constructed to demonstrate the capabilities of these IPMC actuators. This paper further identifies key parameters involving the vibrational and resonance characteristics of sensors and actuators made with IPMCs. When the applied signal frequency varies, so does the displacement up to a critical frequency called the resonant frequency where maximum deformation is observed, beyond which the actuator response is diminished. A data acquisition system was used to measure the parameters involved and record the results in real time basis. Also the load characterizations of the IPMCs were measured and it was shown that these actuators exhibit good force to weight characteristics in the presence of low applied voltages. Finally reported are the cryogenic properties of these muscles for potential utilization in an outer space environment of a few Torrs and temperatures of the order of −140 degrees Celsius. These muscles are shown to work quite well in such harsh cryogenic environments and thus present a great potential as sensors and actuators that can operate at cryogenic temperatures.

1. Introduction

Ion-exchange polymer–metal composites (IPMCs) are active actuators that show large deformation in the presence of low applied voltage and exhibit low impedance. They operate best in a humid environment and can be made as self-contained encapsulated actuators to operate in dry environments as well. They have been modeled as both capacitive and resistive element actuators that behave like biological muscles and provide an attractive means of actuation as artificial muscles for biomechanics and biomimetics applications. Grodzinsky [1], Grodzinsky and Melcher [2, 3] and Yannas and Grodzinsky [4] were the first to present a plausible continuum model for electrochemistry of deformation of charged polyelectrolyte membranes such as collagen or fibrous protein and were among the first to perform the same type of experiment on animal collagen fibers essentially made of charged natural ionic polymers and were able to describe the results through an electro-osmosis phenomenon. Kuhn [5] and Katchalsky [6], Kuhn *et al* [7, 8] and Kuhn and Hargitay [9], however, should be credited as the first investigators to report the ionic chemomechanical deformation of polyelectrolytes such as polyacrylic acid (PAA) and polyvinyl chloride (PVA) systems. Kent *et al* [10] were also the first to report the electrochemical transduction of the PVA–PAA polyelectrolyte system. Recently revived interest in this area concentrates on artificial muscles which can be traced to Shahinpoor and co-workers and other researchers [11–14, 22–53, 57, 62, 63, 66, 70, 71], Osada [15], Oguro *et al* [16], Asaka *et al* [17], Guo *et al* [18] and De Rossi *et al* [19, 20]. More recently De Rossi, Chiarelli, Osada, Hasebe, Oguro, Asaka, Tanaka, Brock, Shahinpoor and Mojarrad [11–71] have been experimenting with various chemically active as well as electrically active ionic polymers and their metal composites as artificial muscle actuators.

Essentially polyelectrolytes possess ionizable groups on their molecular backbone. These ionizable groups have the property of dissociating and attaining a net charge in a variety of solvent media. According to Alexanderowicz and Katchalsky [21] these net charge groups which are attached to networks of macromolecules, called polyions, give rise to intense electric fields of the order of 10^{10} V m^{-1}. Thus, the essence of the electromechanical deformation of such polyelectrolyte systems is their susceptibility to interactions with externally applied fields as well as their own internal field structure. In particular if the interstitial space of a polyelectrolyte network is filled with liquid containing ions, then the electrophoretic migration of such ions inside the structure due to an imposed electric field can also cause the macromolecular network to deform accordingly. Shahinpoor [11, 12, 22–24, 26, 29–32, 34–41, 43, 48, 63, 66, 71, 72] and Shahinpoor and co-workers [13, 14, 25, 27, 28, 33, 42, 44–47, 49–53, 57, 62, 70–72] have recently presented a number of plausible models for micro-electro-mechanics of ionic polymeric gels as electrically controllable artificial muscles in different dynamic environments. The reader is referred to these papers for the theoretical and experimental results on dynamics of ion-exchange membrane–platinum composite artificial muscles.

The IPMC muscle used in our investigation is composed of a perfluorinated ion exchange membrane (IEM), which is chemically composited with a noble metal such as gold or platinum. A typical chemical structure of one of the ionic polymers used in our research is

$$
\begin{array}{l}
[-(\mathrm{CF_2{-}CF_2})_n-(\mathrm{CF{-}CF_2})_m-] \\
\qquad\qquad\qquad\quad | \\
\qquad\qquad\qquad\quad \mathrm{O{-}CF{-}CF_2{-}O{-}CF_2{-}SO_3^-}\ldots\mathrm{M^+} \\
\qquad\qquad\qquad\qquad\quad | \\
\qquad\qquad\qquad\qquad\quad \mathrm{CF_3}
\end{array}
$$

where n is such that $5 < n < 11$ and $m \sim 1$, and M^+ is the counter-ion (H^+, Li^+ or Na^+). One of the interesting properties of this material is its ability to absorb large amounts of polar solvents, i.e. water. Platinum, Pt, metal ions, which are dispersed throughout the hydrophilic regions of the polymer, are subsequently reduced to the corresponding metal atoms. This results in the formation of a dendritic type electrode.

1.1. Metallization of ion-exchange membranes

In metallizing this material there is a first stage of in-depth molecular metallization and a second stage of surface plating and electroding. Thus, the important stage of compositing is the first stage which can be postulated to take place according to the following chemical reactions:

$$\left[\mathrm{Pt}\left(\mathrm{NH_3}\right)_4\right]^{2+} + 2\mathrm{e}^- \rightarrow \mathrm{Pt}^0 + 4\mathrm{NH_3} \tag{1.1}$$

$$\mathrm{LiBH_4} + 8\mathrm{OH}^- \rightarrow \mathrm{BO_2^-} + \mathrm{Li}^+ + 6\mathrm{H_2O} + 8\mathrm{e}^-. \tag{1.2}$$

From equations (1.1) and (1.2), it is possible to have the following reactions:

$$
\begin{aligned}
&\mathrm{LiBH_4} + 4\left[\mathrm{Pt}\left(\mathrm{NH_3}\right)_4\right]^{2+} + 8\mathrm{OH}^- + 8\mathrm{e}^- \rightarrow 4\mathrm{Pt}^0 + 16\mathrm{NH_3} \\
&\quad + \mathrm{BO_2^-} + \mathrm{Li}^+ + 6\mathrm{H_2O} + 8\mathrm{e}^- = \mathrm{LiBH_4} \\
&\quad + 4\left[\mathrm{Pt}\left(\mathrm{NH_3}\right)_4\right]^{2+} + 8\mathrm{OH}^- \rightarrow 4\mathrm{Pt}^0 + 16\mathrm{NH_3} + \mathrm{BO_2^-} \\
&\quad + \mathrm{Li}^+ + 6\mathrm{H_2O}.
\end{aligned} \tag{1.3}
$$

Also, the solid form of $LiBO_2$ occasionally precipitates. Therefore, the overall reaction is most likely,

$$
\begin{aligned}
&\mathrm{LiBH_4} + 4\left[\mathrm{Pt}\left(\mathrm{NH_3}\right)_4\right]^{2+} + 8\mathrm{OH}^- \rightarrow 4\mathrm{Pt}^0(\mathrm{s}) + 16\mathrm{NH_3}(\mathrm{g}) \\
&\quad + \mathrm{LiBO_2}(\mathrm{s}\delta) + 6\mathrm{H_2O}(\mathrm{l}).
\end{aligned} \tag{1.4}
$$

Now, the question of the source of hydroxyl ions may be answered by considering the following reaction to be possible:

$$\mathrm{LiBH_4} + 4\mathrm{H_2O} \rightarrow 4\mathrm{H_2} + \mathrm{LiOH}\left(\mathrm{Li}^+ + \mathrm{OH}^-\right) + \mathrm{B(OH)_3}(\mathrm{s}\delta). \tag{1.5}$$

This indicates 9 moles of $LiBH_4$ are required for reducing 4 moles of $Pt(NH_3)_4^{2+}$.

2. Theoretical consideration

A simple one-dimensional model of electrically induced deformation of ionic polymeric gels is such that

$$\sigma = (1/3)E\left(C_0,\, C_i\right)\left(\lambda - \lambda^{-2}\right) \tag{2.1}$$

$$\sigma = \kappa\left(C_0,\, C_i\right)E^{*2} \tag{2.2}$$

where σ is the stress, λ is the stretch, $E(C_0, C_i)$ is the corresponding Young's modulus of hyper-elasticity, C_0 is the polymer solid concentration, C_i ($i = 1, 2, \ldots, N$) are the molal concentrations of various ionic species in the aqueous medium, $\kappa(C_0, C_i)$ is an electromechanical coefficient and E^* is the local electric field. Thus bending can occur due to differential contraction and expansion of the outermost remote regions of a strip if an electric field is imposed across its thickness as shown in figure 1. Since ionic polyelectrolytes are for the most part a three dimensional network of macromolecules cross-linked nonuniformly, the concentrations of ionic charge groups are also nonuniform within the polymer matrix. Therefore the mechanism of bending is partially related to

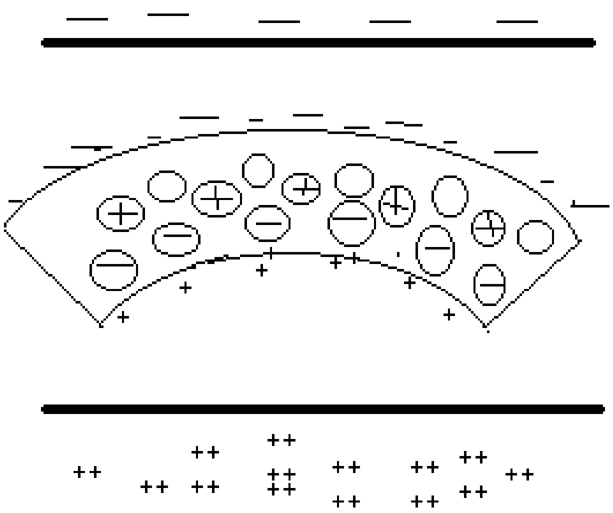


Figure 1. General redistribution of charges in an ionic polymer due to an imposed electric field.

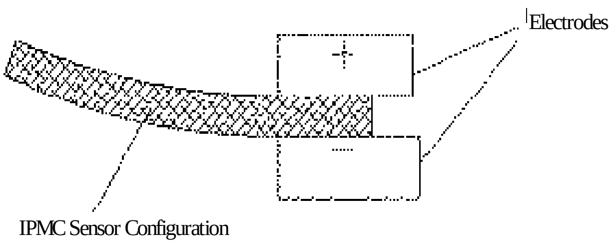


Figure 2. Simple IPMC sensor placed between two electrodes.

the redistribution of fixed ions and migration of mobile ions within the network due to the imposition of an electric field. However, recent modeling effort on the sensing and actuation have revealed that water movement may also play a significant role in actuation. This subject is currently under investigation.

A simple one-dimensional model of electrically induced dynamic deformation or vibration of a cantilever beam made with such IPMC artificial muscle strips is given by the following equations:

$$\rho \frac{\mathrm{d}^2 y}{\mathrm{d}t^2} = \frac{\mathrm{d}\sigma}{\mathrm{d}x} + F(x, t) \tag{2.3}$$

$$\varepsilon = \varepsilon_c + \kappa_E \eta \qquad -C < \eta < C \tag{2.4}$$

$$\lambda = 1 + \varepsilon \tag{2.5}$$

$$\lambda_+ - \lambda_- = 2\kappa_E C \tag{2.6}$$

where F is the body force per unit volume of the muscle, ρ is the density, ε is the strain, subscript c indicates values at the neutral axis of the cross-section of the strip, C is the distance of the outermost remote fibers, κ_E is the local curvature due to an imposed electric field, η is a cross-sectional parameter, E^* is the local electric field, x and t are axial location and time variables and subscripts $+$ and $-$, respectively indicate the values of a variable at the outermost remote fibers.

Thus bending can occur due to differential contraction and expansion of the outermost remote fibers of a strip if an electric field is imposed across its thickness as shown in figures 1 and 2. Numerical solutions to the above set of dynamic equations are presently under way and will be reported later. However, it must be mentioned that the governing equations (2.1)–(2.6) display a set of highly non-linear dynamic equations of motion for the IPMC artificial muscles.

At present attempts are under way to establish the existence and uniqueness of the dynamic solutions to the above equations mathematically. However, experimental observations in our laboratory clearly indicate the non-linear motion characteristics of such muscles as well as unique vibrational response and resonance characteristics.

For detailed dynamics description and analysis of the continuum theory of an ionic polymeric gel the reader is referred to Segalman *et al* [25]. Since polyelectrolytes are for the most part a three-dimensional network of macromolecules cross-linked non-uniformly, the concentrations of ionic charge groups are also non-uniform within the polymer matrix. Therefore the mechanisms of swelling and contraction are intimately related to osmotic diffusion of solvent, ions and counterions into and out of the gel. One possible way to describe this mechanism is to model the system by the governing continuum mechanics equations and neo-Hookean deformation theory. In the next section an analytical relation is presented as described by Segalman *et al* [25].

3. Ion transport mechanisms

Let $c(X, t)$ be the solvent concentration, $H(X, t)$ be the ionic concentration, $x(X, t)$ be the position vector of a typical gel element, X be the reference material coordinate and t be the time such that the governing continuum mechanics equation takes the following forms:

$$\frac{\mathrm{d}c}{\mathrm{d}t} = \Delta\,[\,D_{1,1}(c,\ H)\Delta c + D_{1,2}(c,\ H)\delta H\,] - \Delta(c\dot{x}) \tag{3.1}$$

$$\frac{\mathrm{d}H}{\mathrm{d}t} = \Delta\,[\,D_{2,1}(c,\ H)\Delta c + D_{2,2}(c,\ H)\Delta H\,] - \Delta(H\dot{x}) + \dot{H}_s \tag{3.2}$$

$$\frac{\mathrm{d}\rho_g}{\mathrm{d}t} + \Delta(\rho\dot{x}) = 0 \tag{3.3}$$

$$\rho_g \ddot{x} = \Delta S + \rho_g f_b \tag{3.4}$$

$$\frac{\mathrm{d}\varepsilon}{\mathrm{d}t} = S\Delta\dot{x} + \Delta q + \Delta q_c + J\,E + \rho_g h \tag{3.5}$$

where x is the displacement, a superposed dot stands for a differentiation with respect to time, $D_{i,j}$ is the diffusion coefficient, H_s is the source term for the production of ions in the gel, ρ_g is the gel density, S is the stress tensor, f_b is the body force vector which includes electromagnetic and gravitational terms, ε is the specific internal energy of the ionic polymeric gel, q is the heat flux vector, q_c is the chemical energy flux vector, J is the electric current flux vector, E is the electric field vector and h is the specific source of energy production in the gel. The stress tensor, S, is related to the deformation gradient field by means of

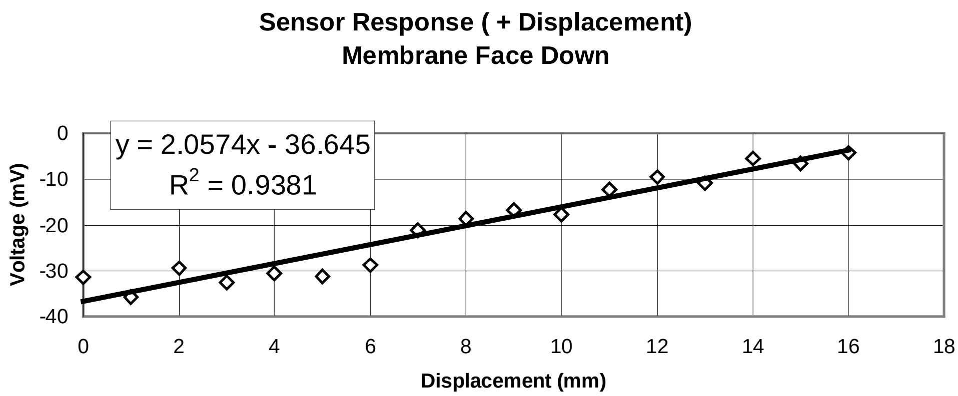


Figure 3. Inverted IPMC film sensor response for positive displacement input.

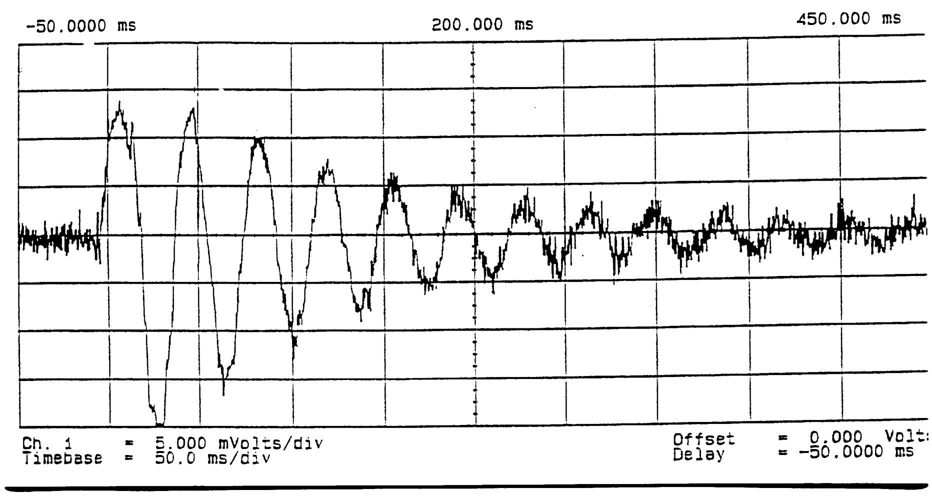


Figure 4. Dynamic sensing response in the form of output voltage of strips (40 mm $\times$ 5 mm $\times$ 0.2 mm) of IPMC subject to a dynamic impact loading as a cantilever.

a neo-Hookean type constitutive equation which may be represented by the following equation:

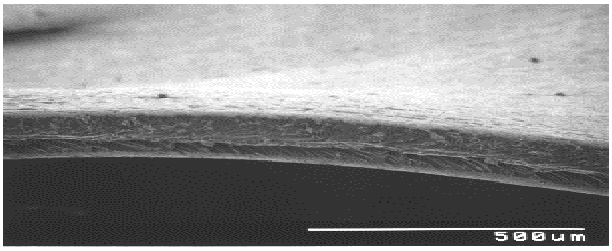
$$S = G(c)\left[F^{-1}\left(F^{-1}\right)^{T} - I\right] + pI \qquad (3.6)$$

where $F = (\partial x/\partial X)$, I is the identity matrix, superscript T stands for transpose, $G(c)$ is the Young's modulus and p is an unknown Lagrangian multiplier to be found by solving the system of equations (2.1)–(2.6) and (3.1)–(3.6). The solution to this model will enable one to electrically control the polymeric muscle bending and therefore the motion of the swimming robotic structure. For additional references on modeling of IPMC artificial muscles the reader is referred to [11–14] and [22–53].

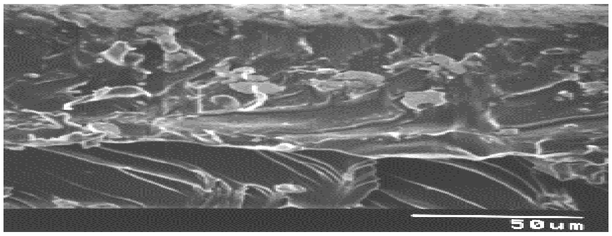
4. Biomimetic sensing capability of IPMC

Investigations of the use of ion-exchange-membrane materials as sensors can be traced to Sadeghipour *et al* [58] where they used such membranes as a pressure sensor/damper in a small chamber which constituted a prototype accelerometer. However, it was Shahinpoor [39] who first discussed the phenomenon of the flexogelectric effect in connection with dynamic sensing of ionic polymeric gels. In this paper the focus is on the application of the IPMC sensor to quasi-static or dynamic displacement sensing where the response of the sensor against large imposed displacements was investigated. To obtain a better understanding of the mechanism of sensing, more explanation must be given of the general nature of the ionic polymers.

As shown in figures 1 and 2, IPMC strips generally bend towards the anode and if the voltage signal is reversed they also reverse their direction of bending. Conversely by bending the material, shifting of mobile charges becomes possible due to imposed stresses. Consider figure 2 where a rectangular strip of the composite sensor is placed between two electrodes. When the composite is bent a stress gradient is built on the outer fibers relative to the neutral axis (NA). The mobile ions therefore will shift toward the favored region where opposite charges are available. The deficit in one charge and excess in the other can be translated into a voltage gradient which is easily sensed by a low power amplifier.



(a)



(b)

Figure 5. Scanning electron micrographs of the structure of IPMC: (a) displays the thickness edge of the muscle while (b) depicts the metal particle deposition on the network inside the muscle.

4.1. Quasi-static sensing

The experimental results showed that a linear relationship exists between the voltage output and imposed quasi-static displacement of the tip of the IPMC sensor as shown in figure 3. The experimental set-up was such that the tip of the cantilevered IPMC strip as shown in figure 2 was mechanically moved and the corresponding output voltage recorded. The results are shown in figure 3.

4.2. Dynamic sensing

When strips of IPMC are dynamically disturbed by means of a dynamic impact or shock loading, a damped electrical response is observed as shown in figure 4. The dynamic response was observed to be highly repeatable with a fairly high band width to hundreds of Hz. This particular property of IPMCs may find a large number of applications in large motion sensing devices for a variety of industrial applications. Since these muscles can also be cut as small as one desires, they present a tremendous potential for micro-electro-mechanical systems (MEMS) sensing and actuation applications.

5. Biomimetic actuation properties of IPMCs

5.1. General considerations

As mentioned before, IPMCs are large motion actuators that operate under a low voltage compared to other actuators such as peizoceramics or shape memory alloys. Table 1 shows a comparison between the capability of IPMC materials and both electroceramics and shape memory alloys. As shown in table 1, IPMC materials are lighter and their potential striction capability can be as high as two orders of magnitude more than EAC materials.

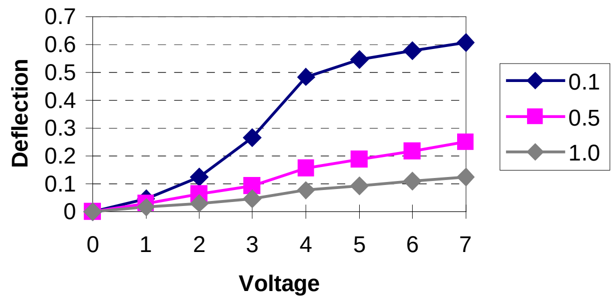


Figure 6. The deflection of an IPMC strip as a function of the frequency (0.1, 0.5 and 1 Hz) and the applied voltage.

Further, their response time is significantly higher than shape memory alloys (SMAs). They can be designed to emulate the operation of biological muscles and have unique characteristics of low density as well as high toughness, large actuation strain and inherent vibration damping.

These muscles are manufactured by a unique chemical process in which a noble metal (Pt) is deposited within the molecular network of the base ionic polymer. Equations (1.1) through (1.5) depict the essence of such chemical compositing which is followed by a surface plating and electroding process. One of the interesting properties of IPMC artificial muscle is its ability to absorb large amounts of polar solvents, i.e. water. Platinum salt ions, which are dispersed throughout the hydrophilic regions of the polymer, are subsequently chemically reduced to the corresponding metal atoms. This results in the formation of dendritic type electrodes. In figure 5, scanning electron micrographs are shown in two magnifications, with an order of magnitude difference. In (a), a view is given of the edge of an electroded muscle. The Pt metal covers each surface of the film with some of the metal penetrating the subsurface regions of the material. A closer view with $\times 10$ magnification is shown in figure 5(b).

When an external direct voltage of 2 volts or higher is applied to an IPMC film, it bends towards the anode. An increase in the voltage level (up to 6 or 7 volts) causes a larger bending displacement. When an alternating voltage is applied, the film undergoes swinging movement and the displacement level depends not only on the voltage magnitude but also on the frequency. Lower frequencies (down to 0.1 or 0.01 Hz) lead to higher displacement (approaching 25 mm) for a 0.5 cm $\times$ 2 cm $\times$ 0.2 mm thick strip. Thus, the movement of the muscle is fully controllable by the applied electrical source. The muscle performance is also strongly dependent on the water content which serves as an ion transport medium and the dehydration rate gradient across the film leads to a pressure difference. The frequency dependence of the ionomer deflection as a function of the applied voltage is shown in figure 6. A single film was used to emulate a miniature bending arm that lifted a mass weighing a fraction of a gram. A film-pair weighing 0.2 g was configured as a linear actuator and using 5 V and 20 mW successfully induced more than 11% contraction displacement. Also, the film-pair displayed a significant expansion capability, where a

Table 1. Comparison of the properties of IPMC, SMA and EAC.

Property	Ionic polymer–metal composites (IPMC)	Shape memory alloys (SMA)	Electroactive ceramics (EAC)
Actuation displacement	>10%	<8% short fatigue life	0.1–0.3%
Force (MPa)	10–30	about 700	30–40
Reaction speed	μs to s	s to min	μs to s
Density	1–2.5 g cm^{-3}	5–6 g cm^{-3}	6–8 g cm^{-3}
Drive voltage	4–7 V	NA	50–800 V
Power consumption	watts	watts	watts
Fracture toughness	resilient, elastic	elastic	fragile

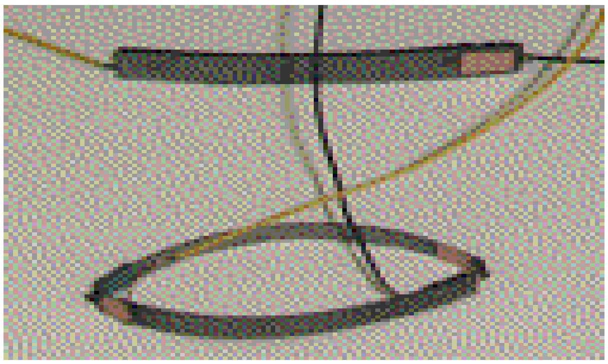


Figure 7. IPMC film-pair in expanded mode. A reference pair (top) and an activated pair (bottom).

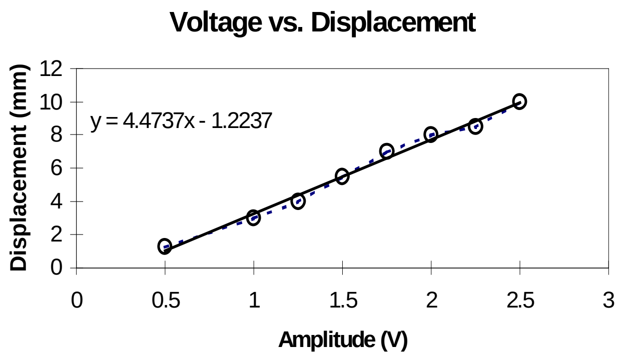


Figure 9. Bending displacement versus voltage for a typical IPMC strip of 5 mm × 0.20 mm × 20 mm under a frequency of 0.5 Hz.

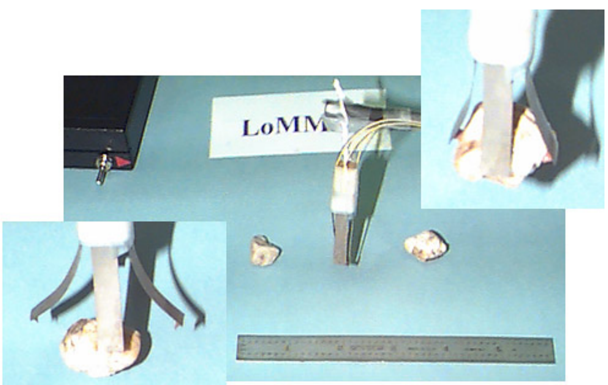


Figure 8. An end-effector gripper lifting a 10.3 g rock under 5 V, 25 mW activation using four 0.1 g fingers made of IPMCs.

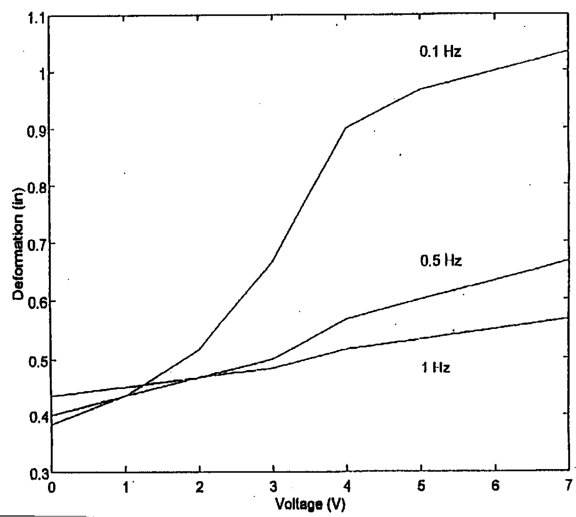


Figure 10. Frequency dependence of bending deformation of IPMC composite muscles.

stack of two film-pairs 0.2 cm thick expanded to about 2.5 cm wide (see figure 7).

5.2. Muscle actuators for soft robotic applications

IPMC films have shown remarkable displacement under relatively low voltage, using very low power. Since the IPMC films are made of a relatively strong material with a large displacement capability, we investigated their application to emulate fingers. In figure 8, a gripper is shown that uses IPMC fingers in the form of an end-effector of a miniature low-mass robotic arm.

The fingers are shown as vertical gray bars and the electrical wiring, where the films are connected back-to-back, can be seen in the middle portion of figure 8. Upon electrical activation, this wiring configuration allows the fingers to bend either inward or outward similar to the operation of a hand and thus close or open the gripper fingers as desired. The hooks at the end of the fingers represent the concept of nails and secure the gripped object that is encircled by the fingers.

To date, multi-finger grippers that consist of two and

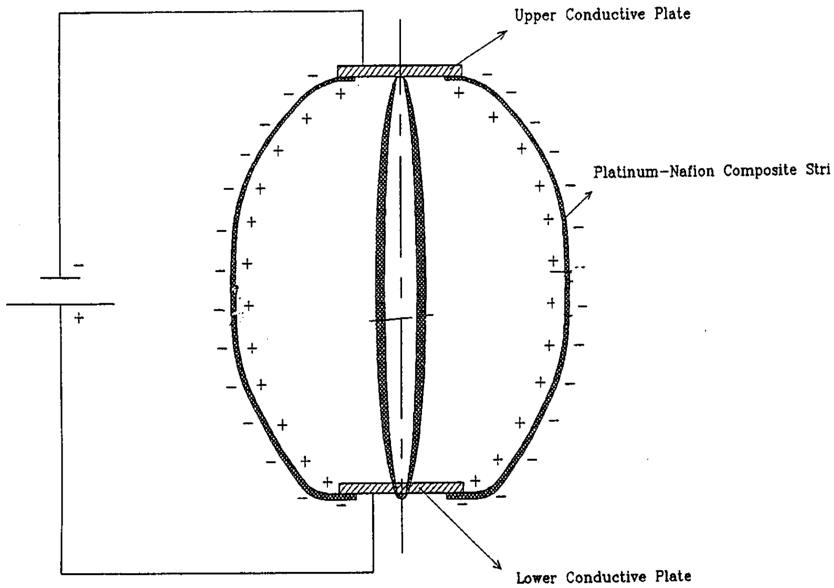


Figure 11. A typical linear-type robotic actuator made with IPMC legs.

four fingers have been produced, where the four-finger gripper shown in figure 8 was able to lift 10.3 g. This gripper prototype was mounted on a 5 mm diameter graphite/epoxy composite rod to emulate a light-weight robotic arm. This gripper was driven by a 5 volts square wave signal at a frequency of 0.1 Hz to allow sufficient time to perform a desirable demonstration of the capability of the gripper—opening the gripper fingers, bringing the gripper near the collected object, closing the fingers and lifting an object with the arm. The demonstration of this gripper capability to lift a rock was intended to pave the way for a future potential application of the gripper to planetary sample collection tasks (such as Mars exploration) using an ultra-dextrous and versatile end-effector.

5.3. Linear and platform type actuators

For detailed dynamics description and analysis of the dynamic theory of ionic polymeric gels the reader is referred to Shahinpoor and co-workers [11–14, 22–70]. Since polyelectrolytes are for the most part a three-dimensional network of macromolecules cross-linked non-uniformly, the concentrations of ionic charge groups are also non-uniform within the polymer matrix. Therefore the mechanism of bending is partially related to migration of mobile ions within the network due to imposition of an electric field as shown in figure 1. However, recent investigation by the author and his co-workers point to a stronger effect due to surface charge interactions which will be reported later. Figure 9 depicts the bending deformation of a typical strip with varying electric field, while figure 10 displays the variation of deformation with varying frequency of alternating electric field.

Based on such dynamic deformation characteristics, linear and platform type actuator can be designed and made dynamically operational. These types of actuators are shown in figures 11 and 12.

6. Large amplitude vibrational response of IPMCs

6.1. General considerations

Strips of IPMC were used to study their large amplitude vibration characteristics. The IPMC strips were chemically composited with platinum. A small function generator circuit was designed and built to produce an approximately ± 4.0 V amplitude alternating wave at varying frequency. In order to study the feasibility of using IPMC artificial muscles as vibration dampers, a series of muscles made from IPMCs were cut into strips and attached either end-to-end or to one fixed platform and another movable platform in a cantilever configuration. By applying a low voltage the movement of the free end of the beam could be calibrated and its response measured, accordingly. Typical data for the frequency dependence of the amplitude of lateral oscillations of the muscle strips subjected to alternating voltages of various forms such as sinusoidal, rectangular, saw-tooth or pulsed were investigated. Furthermore, the static deformation of the strip with voltage as well as the frequency dependence of deflection–voltage curves were evaluated.

6.2. Experimental observations

IPMC artificial muscle strips of about 2–4 cm $\times$ 4–6 mm were cut and completely swollen in a suitable solution such as water. The IPMC muscle strip typically weighed 0.1–0.4 grams and its thickness measured about 0.2 mm. The strip was then held by a clamping set-up between two platinum plate terminals which were wired to a signal amplifier and generator apparatus driven by Labview software through an IBM compatible PC containing an analog output data acquisition board. The amplifier (Crown model D-150A) was used to amplify the signal output of a National Instrument data acquisition card (AT-AO-10). Software was written to produce various waveforms such

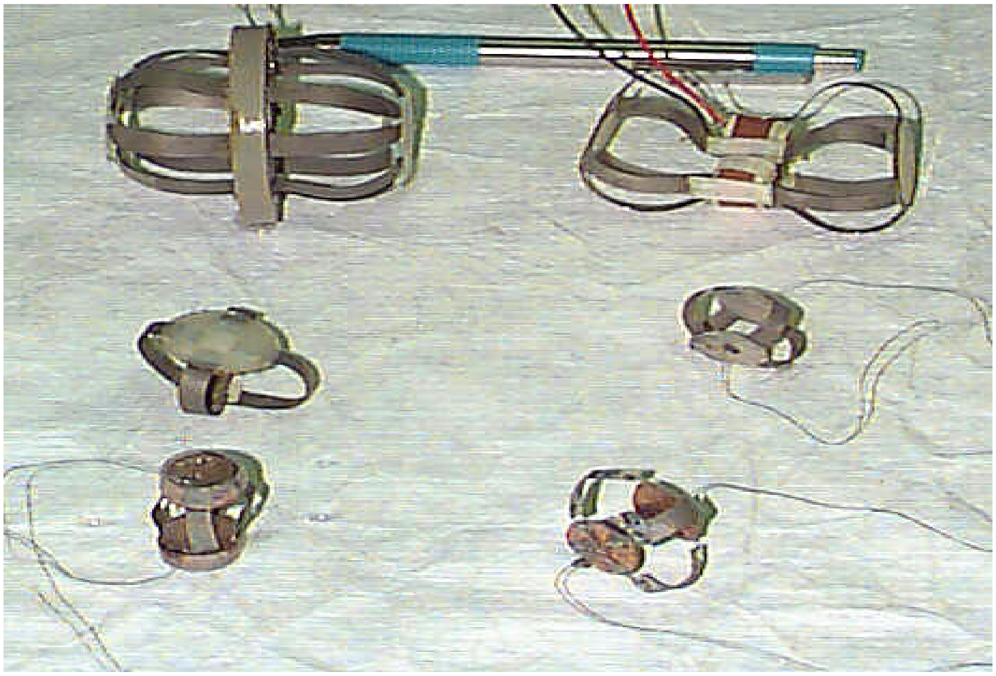


Figure 12. An assortment of linear and platform type actuators based on the design depicted in figure 11.

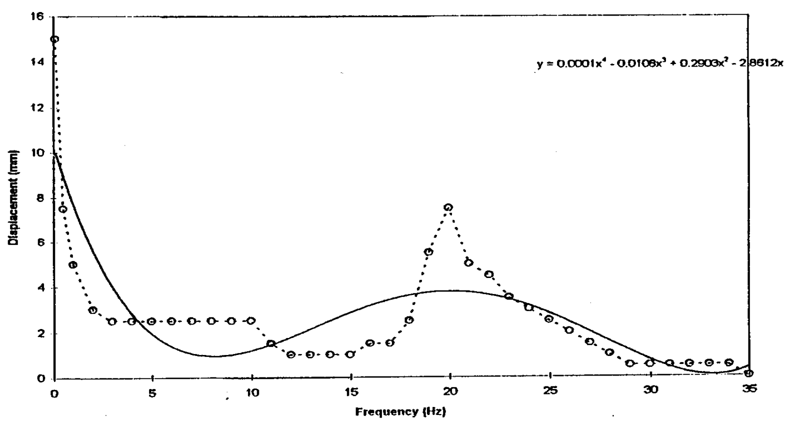


Figure 13. Amplitude of displacement versus the imposed frequency for a voltage of 2 volts for a 2 cm $\times$ 4 mm $\times$ 0.2 mm sample.

as sinusoid, square, triangular and saw tooth signals at desired frequencies up to 100 Hz and amplitudes up to 10 volts. When a low direct voltage was applied, the membrane composite bent toward the anode side each time. So by applying an alternating signal we were able to observe alternating bending of the actuator that followed the input signal very closely up to 35 Hz. At voltages higher than 2.0 volts, degradation of displacement output of the actuator was observed which may be due to dehydration. Water acts as the single most important element for the composite bending by sequentially moving within the composite depending on the polarity of the electrodes. The side facing the anode dehydrated faster than the side facing the cathode leading to a differential stresses which ultimately leads to bending of the composite. So, prior to each experiment, the composite was completely swollen in water. The displacement of the free end of a typical 2 cm $\times$ 4 mm composite membrane was then measured for the frequency range of 0.1–35 Hz for sinusoid input voltage at 2.0 volts amplitude (figure 13).

Resonance was observed at about 20 Hz where the associated displacement was observed to be 7.5 mm. It should be noted that as the actuator dehydrated the resonance frequency and maximum displacement varied accordingly. By encapsulating the strips in a plastic membrane such as Saran^{R}, the deterioration in the amplitude of oscillation decreased with time. However, the initial amplitude of oscillation for the same level of voltage was smaller than the unwrapped case due to increased rigidity of the strip. For our sample actuator the resonance occurred in the frequency range of 12 to 28 Hz.

Based on such dynamic deformation characteristics, noiseless swimming robotic structures as shown in figure 14 and cilia assembly-type robotic worlds, similar to coral

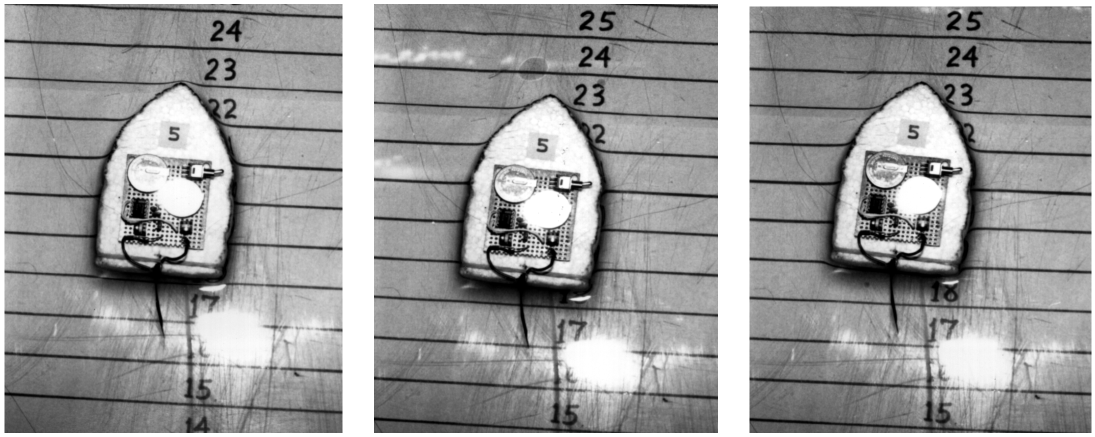

(a) (b) (c)

Figure 14. Robotic swimmer with muscle undulation frequency of 3 Hz (frame time interval, 1/3 second).

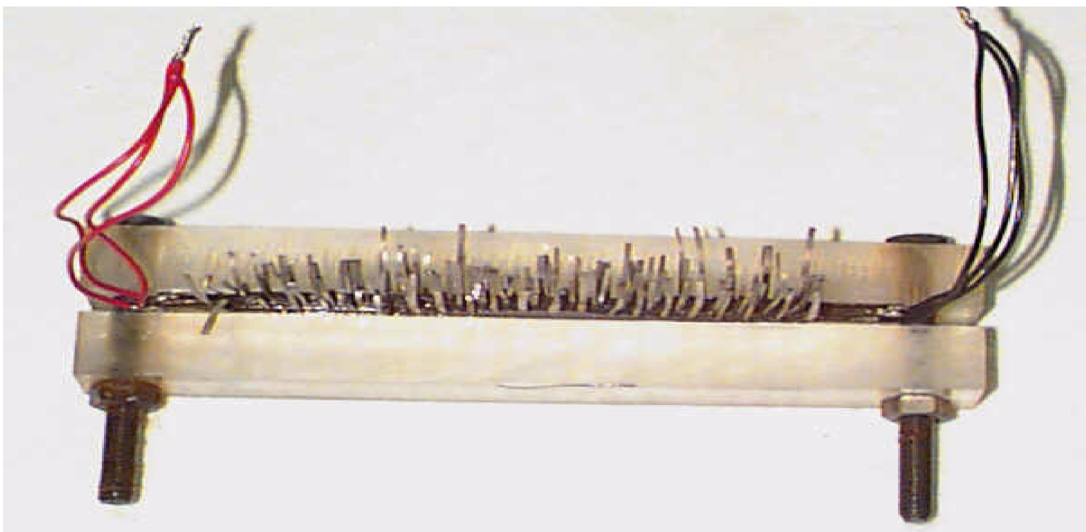


Figure 15. Cilia-type assembly of IPMC–Pt muscles simulating collective dynamic vibrational response similar to coral reefs could create anti-biofouling surfaces.

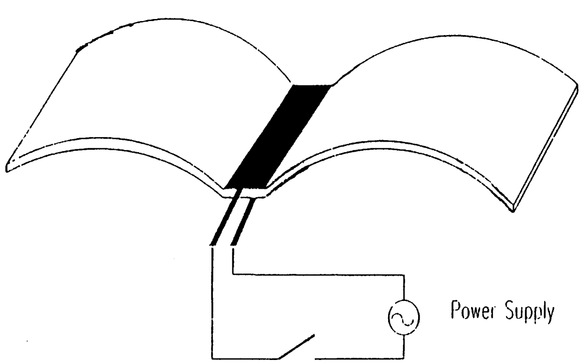


Figure 16. Wing-flapping flying machine design depicted schematically.

reefs, as shown in figure 15, were constructed and tested for collective vibrational dynamics. Furthermore, wing flapping flying machines, schematically shown in figures 16 and 17 can be equipped with these muscles.

7. Load and force characterization of IPMCs

7.1. General considerations

In order to measure the force generated by strips of these muscles in a cantilever form an experimental set-up was designed using a load cell. A load cell (Transducer Techniques, model GS-30, 30 grams capacity) and corresponding signal conditioning module (Transducer Techniques, model TMO-1) together with a power supply was set up and connected to a PC-platform data acquisition and signal generation system composed of a 12-bit analog output board (National Instrument AT-AO-10) and a 16-bit multi-input–output board (National Instrument AT-MIO-16XE-50). A Nicolet scope was used to monitor the input and output waveform. Labview^{TM} software was used to write a program to generate various waveforms such as sinusoid, square, saw tooth and triangular signals at desired frequencies and amplitudes. The effective length of the membrane was 10 mm. This made the effective weight

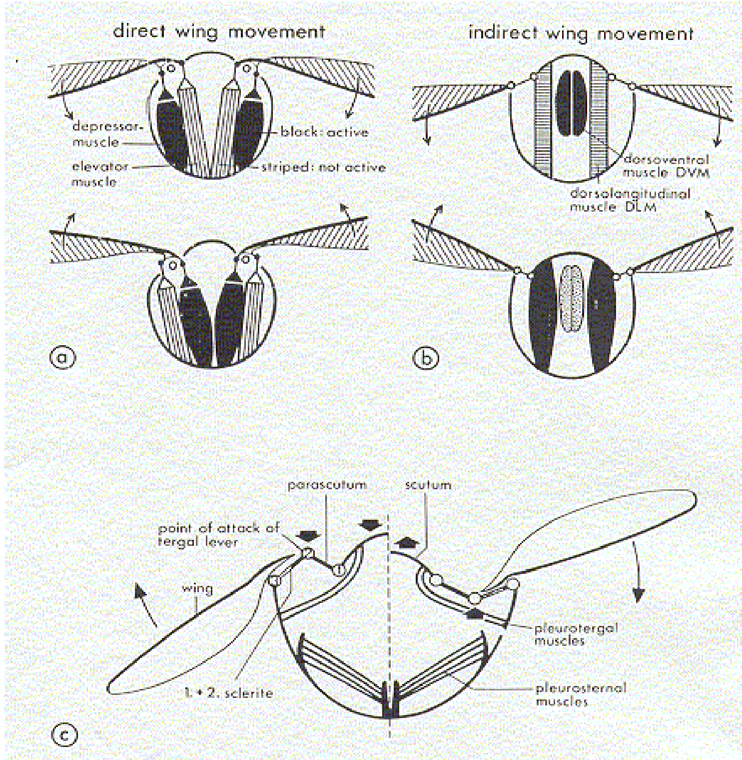


Figure 17. A simple design similar to early flight mechanisms for insects, in which linear type actuators as depicted in figures 11 and 12 can be used to operate simple flying machines, (a) direct wing movement, (b) indirect wing movement, (c) asymmetric form.

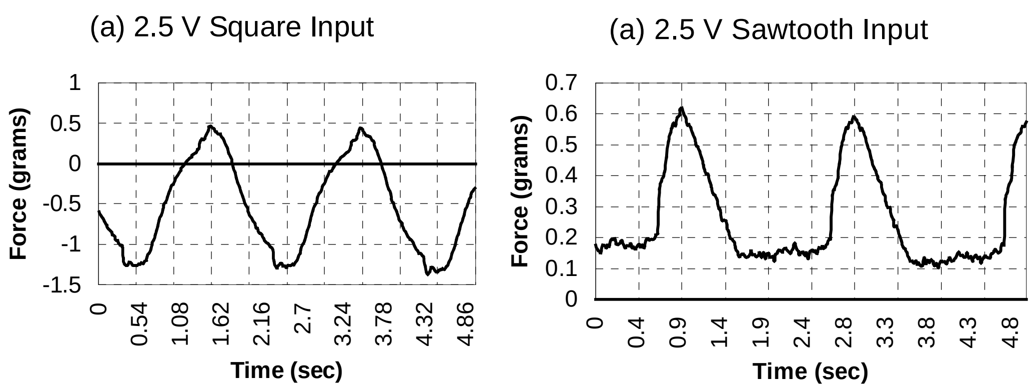


Figure 18. IPMC actuator response for square and saw tooth wave input at 2.5 V rms and a current of about 20 milliamps.

of the muscle producing a force about 20 milligrams. The resulting graphs were then adjusted for initial noise and preload and plotted over a 5 second period (2.5 cycles). The force capability of these muscles, on average, was measured to be about 400 N kg m^{-1} indicating that these muscles can lift almost 40 times their own weight. Figures 18 depicts such general trends.

7.2. Force improvement by means of chemical tweaking

It has recently been established that with the tweaking of the chemical composition of IPMC the force capability of these muscles can be greatly improved. This is one of the most important results of the present article. Reported here is a set of data pertaining to force capability and recent force improvement on ionic polymeric platinum composite muscles. The experimental configuration of the muscle when we measure the force, by a force transducer, is in a cantilever form as shown in figure 2. A voltage is applied to one end by a pair of electrodes and the force against the transducer on the other end is measured. The form of the voltage signal is also important. The best results seem to be produced by a square type of voltage signal. Typically for a square voltage signal of 3 volts maximum, and 40 milliamps, a muscle strip of 5 mm $\times$ 20 mm $\times$ 0.2 mm weighing about 30 milligrams produces a completely reversible force of about 1.2 grams

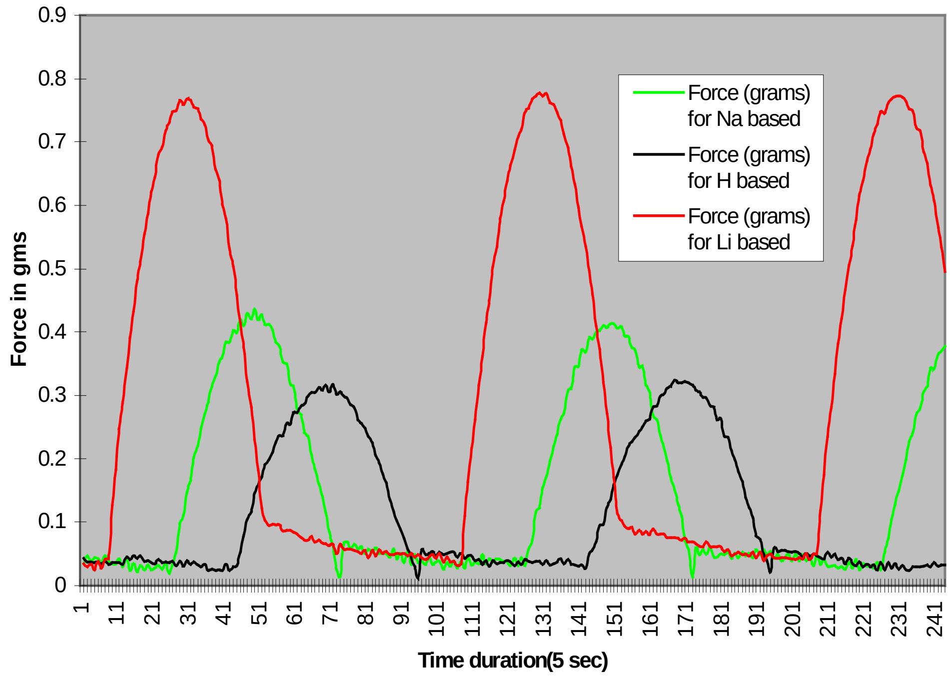


Figure 19. Force improvement by a factor of 2, by means of chemical tweaking.
(This figure can be viewed in colour in the electronic version of the article; see `http://www.iop.org`)

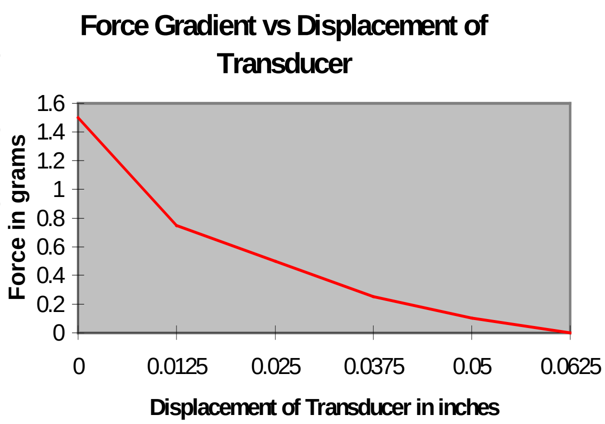


Figure 20. Tip force versus tip displacement of the cantilever muscle strip (lithium–cation-based muscle).

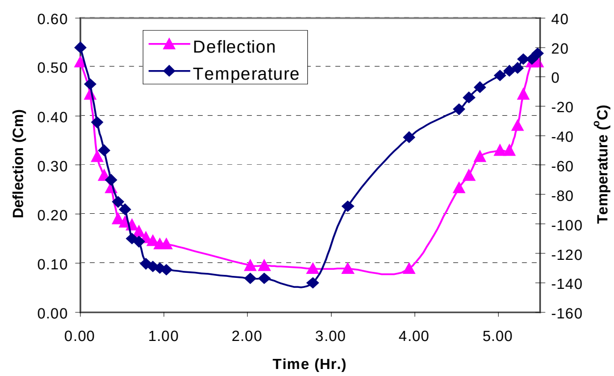


Figure 21. Deflection characteristics of IPMC as a function of time and temperature.

at a frequency of 0.5 Hz. This means a force density of about 400 N kg^{-1} m^{-1}, which roughly means these muscles could lift 40 times their own weight. Figure 19 depicts the variation of force under a sinusoidal voltage signal in a cantilever configuration in which the muscle strip only comes into contact with a force transducer over half the cycle.

7.3. Force versus displacement

The variations of tip force versus tip displacements in a cantilever configuration was obtained using muscles which were lithium-based muscle strips of 1 in by 1/4 in dimensions and 0.2 mm thick. These muscle strips were tested under a voltage of 2 volts at a frequency of 0.5 Hz in a cantilever configuration. The results are shown in figure 20. The results indicate that the behaviour is similar to a piecewise linear spring with large deformation.

8. Cryogenic properties of IPMC artificial muscles

In this section are reported a number of recent experimental results pertaining to the behavior of ionic polymer metal composites (IPMCs) under low pressure (few Torrs) and low temperature (−140 degrees Celsius). These experimental results have been obtained in a cryogenic chamber at NASA/JPL as well as a cryogenic chamber at the Artificial Muscles Research Institute at UNM. The interest at NASA/JPL was to study the actuation properties of these muscles in a harsh space environment such as one Torr of pressure and −140 degrees Celsius temperature.

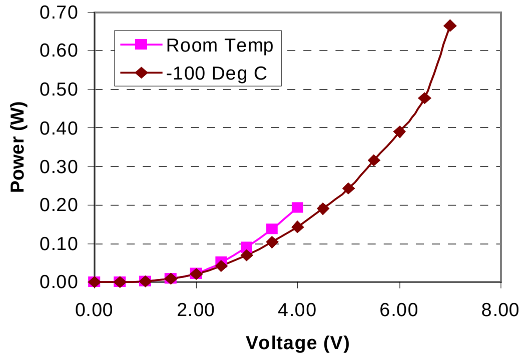


Figure 22. Power consumption of the IPMC strip bending actuator as a function of activation voltage.

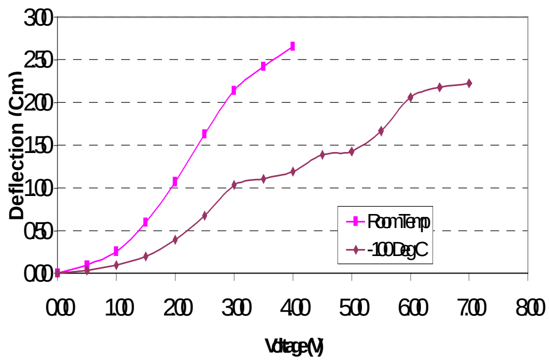


Figure 23. Deflection of the bending IPMC strip as a function of voltage.

While at UNM the electrical properties, sensing capabilities as well as actuation properties of these muscles were tested in an atmospheric pressure chamber with a low temperature of −80 degrees Celsius.

In general the results show that these materials are still capable of sensing and actuation in such harsh conditions as figures 21 through 28 display. Furthermore, these IPMC artificial muscles become less conductive, i.e., their electrical resistance increases with decreasing temperature. This result appears to defy the generally accepted fact that resistance of metallic conductors increases/decreases with increasing/decreasing temperature, respectively.

Figure 25(a) clearly shows a remarkable trend which is opposite to the normal trend of resistance–temperature variations in conductors. The graph shows that as the temperature decreases in IPMC artificial muscles the resistance increases. For any given temperature, there is a range of linear response of V against I, which indicates a close to a pure resistor response. This rather remarkable effect is presently under study. However, one plausible explanation is that the lower the temperature the less active are the ionic species within the network of the IPMC and thus the less the ionic current activities. Since current is voltage over the resistance R, i.e., $I = V/R$, thus R has to increase to accommodate the decreasing ionic current due to decreasing temperature.

Figures 26, 27 and 28 show the relationship between the temperature, voltage, current, power and displacement in typical IPMC strips. Note that the behavior of this material

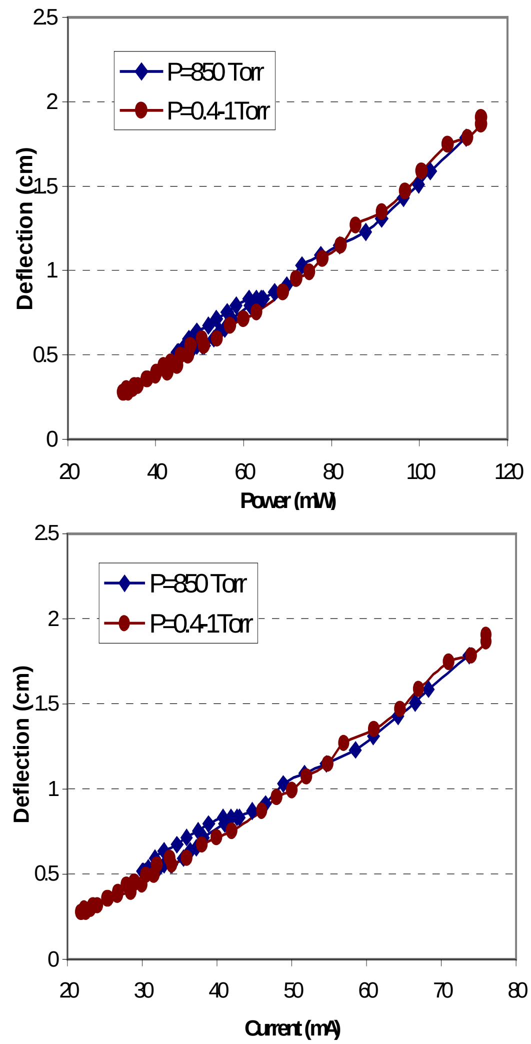


Figure 24. Deflection versus power and current under a constant voltage of 3 volts and a frequency of 0.1 Hz for two different pressures.

at low temperatures resembles more a semi-conductor type response to colder temperatures rather than a typical metallic conductor.

9. Summary

An introduction to ionic polymer metal composites as biomimetic sensors and actuators was presented. Some theoretical modeling on the mechanisms of sensing and actuation of such polymer composites was given. Highly dynamic sensing characteristics of IPMC strips were remarkable in accuracy and repeatability and found to be superior to existing motion sensors and micro-sensors. A new type of soft actuator and multi-fingered robotic hand were made from IPMC artificial muscles and found to be quite superior to conventional grippers and multi-fingered robotic hands. The feasibility of designing linear and platform type robotic actuators made with IPMC artificial muscle was presented. By applying a low voltage the movement of the free end of the actuator could be calibrated

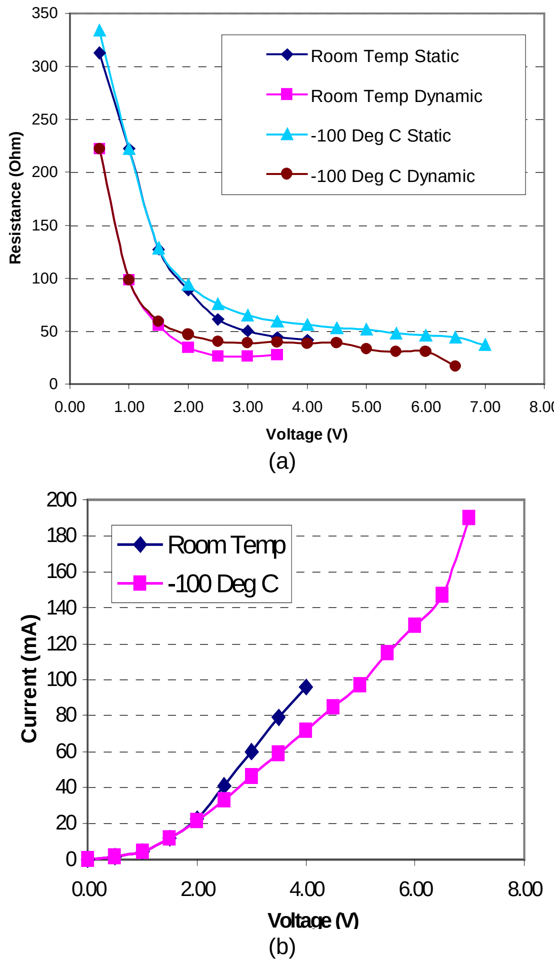


Figure 25. Effect of temperature on the electrical resistance. (a) IPMC strip static (V/I) and dynamic (V/I) resistance at two temperatures. (b) The relation between voltage and current for an IPMC strip that was exposed to RT and to $-100\,^{\circ}$C.

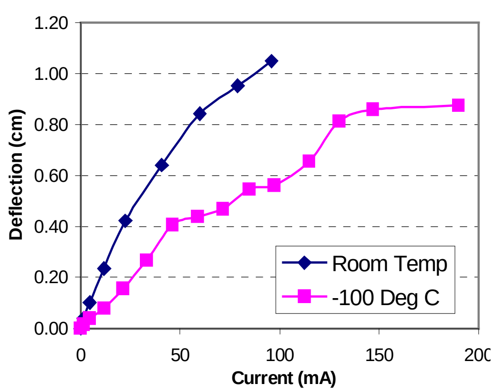


Figure 26. The relation between the current and the deflection for an IPMC strip that was exposed to room temperature and to $-100\,^{\circ}$C.

and its response could be measured, accordingly. The feasibility of designing dynamic vibrational systems of artificial muscles made with IPMC artificial muscle were presented. Our experiments confirmed that these types of composite muscle show remarkable bending displacement

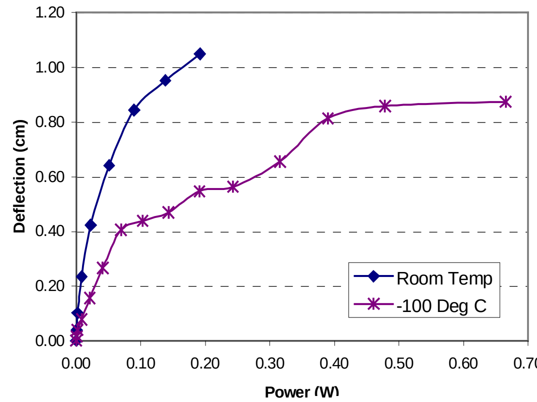


Figure 27. The relation between the power and the deflection for an IPMC strip that was exposed to room temperature and to $-100\,^{\circ}$C.

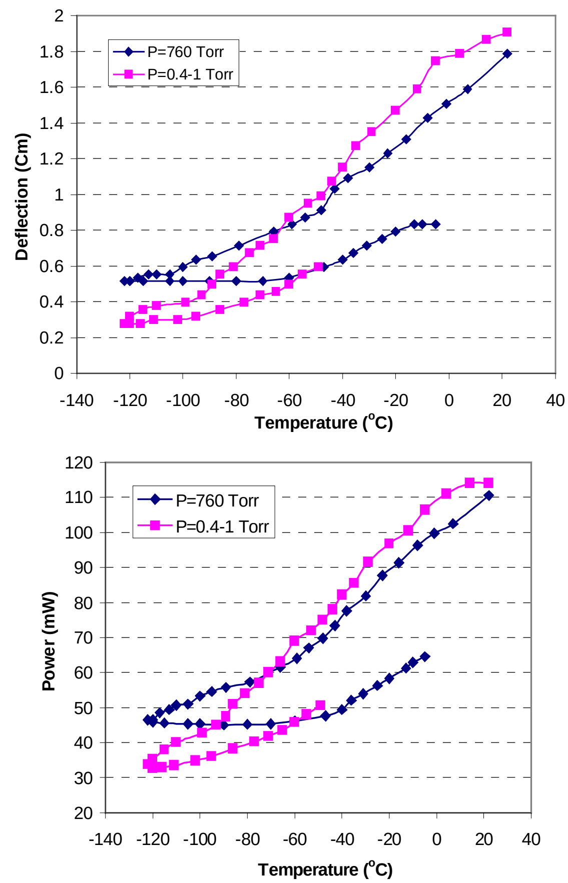


Figure 28. Deflection and power consumption of the IPMC muscle as a function of temperature with pressure as a parameter. V_{peak} = 3 V, frequency = 0.1 Hz.

that follow the input signal very closely. When the applied signal frequency varies, so does the displacement up to a point where large deformations are observed at a critical frequency called the resonant frequency where maximum deformation was observed, beyond which the actuator response was diminished. A data acquisition system was used to measure the parameters involved and record the results in real time basis. The observed remarkable vibrational characteristics of IPMC composite artificial muscles clearly point to the potential of these muscles for biomimetics applications such as swimming robotic structures, wing-flapping flying machines, slithering snakes, heart and circulation assist devices, peristaltic pumps and dynamic robotic cilia-worlds. The cryogenic properties of these materials were quite unique. The fact that they still operated at very low temperatures such as -140 degrees Celsius shows their potential as cryogenic sensors and actuators. Their resistance increased with decreasing temperature, a property that is opposite to all metallic conductors.

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