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Editorial

Soft Biomimetic Robotic Looped Haptic Feedback Sensors

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Abstract

Reported is a new family of Ionic Polymer Metal Composites (IPMCs) sensors and actuators in the form of a loop, made by bending a strip of IPMC around to make an end-to-end cross and form a looped haptic robotic feedback sensor and actuator. The looped cantilever IPMCs is experimentally shown to be capable of bending and twisting as well as sensing characteristics. The sensing characteristics as a soft biomimetic haptic feedback sensor are shown to have great potential for ubiquitous Robot-Human Interactions (RHI) as well as providing haptic force feedback, for example, to surgeons during robotic surgery. Upon various types of deformations, they are shown to generate unique output voltage signal and transient current to be correlated to the actual haptic feedback force. Furthermore, the looped IPMC haptic feedback sensor can enable a new advanced technology in robotic surgery by providing surgeons efficient routines for kinesthetic and softness inquiry of organs and tissues during robotic surgery and yet they can be actuated simultaneously on the fly by a small voltage (4-6 volts) for reconfiguration of looped IPMC feedback sensors for normal grasping and manipulation of bodily organs and tissues. Also, looped IPMC haptic force feedback sensors can be applied as a smart skin for the development of human-like dexterous and soft manipulation.

Introduction

Ionic Polymer-Metal Composites (IPMCs) belong to a family of electro active polymers that deform spectacularly (in actuation mode) by a small imposed electric field (few kV/m) and also generate electrical fields (sensing and energy harvesting mode) upon physically deforming them (mechanically or by environmental dynamics, such as wind or other natural pulses). They work both in the air and in polar liquids such as water and blood. Ionic Polymer-Metal Composites (IPMCs) are synthetic composite nano materials that display artificial muscle behavior under an applied voltage or an electric field. IPMCs are composed of an ionic polymer like Nafion® or Flemion® whose surfaces are chemically plated or physically coated with conductors such as platinum or gold. Under an applied voltage (1-4 V for typical samples of the size 10mmx40mmx0.2mm), ion migration and redistribution due to the imposed voltage across a strip of IPMC result in bending deformation. If the plated electrodes are arranged in a non-symmetric configuration, the imposed voltage can induce another type of deformations like twisting, rolling, turning, twirling, whirling, and non-symmetric bending deformation. Alternatively, if such deformations are physically applied to the IPMC strips they generate an output voltage signal (few mill volts for typical small samples (5mmx30mmx0.2mm)) as sensors and energy harvesters (Figure 1, Figure 2). They have a force density of about 40 in a cantilever configuration with sizes of 5mmx30mmx0.2mm, meaning that they can generate a tip blocking force of almost 40 times their weight in a cantilever mode. In this case, the weight of the cantilever is around 0.06 gmf based on a density of 2 gm/cm^3 for IPMCs which means it can produce a tip blocking force of 2.4 gmf. Another character of these type samples of IPMCs in actuation, sensing, and energy harvesting modes is to have a very broad bandwidth up to kilo HZ and higher. IPMC was introduced in 1998 by Shahinpoor, Bar-Cohen, Xue, Simpson, and Smith [1,2]. However, the original idea of ionic polymer actuators and sensors goes back to 1992- 93 results by Osada, et al. [3], Oguro, et al. [4-6], Segalman, et al. [7], Shahinpoor [8], Doi, et. al [9] and Adolf, et al. [10]. For manufacturing 3D samples of IPMCs refer to [11-14].

The essential mechanism for both actuation and sensing/energy harvesting capabilities of IPMCs is the migration of cations (Na^+, Li^+), which are loosely adjoined to the underlying molecular network with anions, towards the cathode electrode and away from the anode electrode due to either an imposed electric field (actuation) or an imposed deformation field (sensing/ energy harvesting). (Figure 1) displays the actuation and sensing mechanisms in cantilever strips of IPMCs in a graphical manner. For modeling of IPMC's actuation, energy harvesting, and sensing see references [15-22].

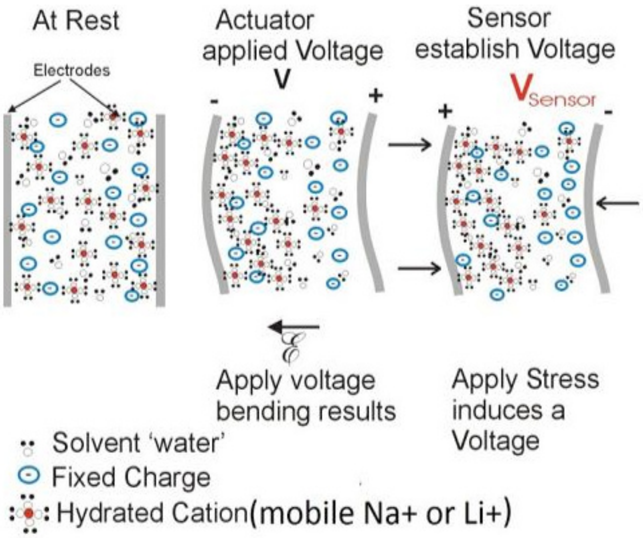


Figure 1: Essential mechanism of actuation and sensing in IPMCs.

IPMC Modeling and Simulation

de Genes and coworkers [15] presented the first phenomenological theory for sensing and actuation in ionic polymer metal composites. Asoka, et al., [16] discussed the bending of polyelectrolyte membrane-platinum composites by electric stimuli and presented a theory on actuation mechanisms in IPMC by considering the electro-osmotic drag term in transport equations. The underlying principles of the Ionic polymeric nano composites' actuation and sensing capabilities can be described by the standard Onsager formulation using linear irreversible thermodynamics modeling. When static conditions are imposed, a simple description of mechanoelectric effect is possible based upon two forms of transport: ion transport (with a current density, normal to the material) and solvent transport (with a flux, we can assume that this term is water flux). The conjugate forces include the electric field and the pressure gradient, ▼the resulting equation has the concise form of.

$$J\,(x, y, z, t) = \sigma\, E_e\,(x, y, z, t) - L_{11}\nabla p\,(x, y, z, t) \quad (1)$$
$$Q\,(x, y, z, t) = L_{21}E\,(x, y, z, t) - K\nabla p\,(x, y, z, t) \quad (2)$$

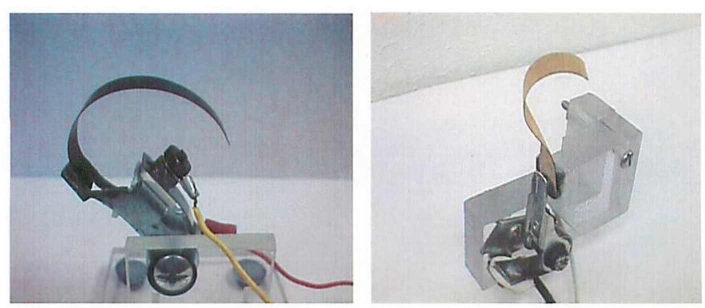
Where, and K are the material electric conductance and the Darcy permeability, respectively. A cross coefficient is usually $L = L_{12} = L_{21}$. The simplicity of equations 1 and 2 provides a compact view of the underlying principles of actuation, transduction, and sensing of the ionic polymer nano composites. When measuring the direct effect (actuation mode) by imposing an electric field in the presence of electrodes which are impermeable to ion species flux, = 0. This gives:

$$\nabla p\,(x, y, z, t) = \frac{L}{K} E_e\,(x, y, z, t) \quad (3)$$

The $\blacktriangledown_p$ (x, y, z, t) will, in turn, induce a k curvature proportional to $\blacktriangledown_p$ (x, y, z, t)The relation between the curvature k and the pressure gradient $\blacktriangledown_p$ (x, y, z, t) has been derived and described in de Genes, Okumura, Shahinpoor and Kim [15].

The experimental deformation characteristics of IPMCs [17-21] are consistent with the above predictions obtained by the above linear irreversible thermodynamics formulation has been used to estimate the value of the Onsager coefficient L to be of the order of 10^{-8} m^2/V-s. Other parameters have been experimentally measured to be K~10^{-18} m^2/CP, σ~1A/mV or S/m. Also, regarding more details on charge transport modeling of actuation and sensing (Poisson-Nernst-Planck equations) refer to Bahramzadeh and Shahinpoor [20-21] and Shahinpoor [22].

In the actuation mode, the cations migrate towards the cathode when an electric field is applied across the IPMC strip. Bending occurs towards the anode side because the anode side is in contraction due to depletion of cations and the cathode side is in tension due to the arrival of cations. This scenario forces the strip to bend accordingly. If the electrodes placed non-symmetrically on the IPMC strip, it can bend, and twist as also shown in (Figure 2a and 2b).



(a) **(b)**

Figure 2: Deformation (bending (a), bending & twisting (b)) of typical small strips of IPMCs (4cmx1cmx0.2mm) in a small electric field of 20kV/m.

(Figure 3 and 4) depict typical test result of deformation of IPMC strips in a cantilever configuration versus voltage and current both in actuation and sensing modes.

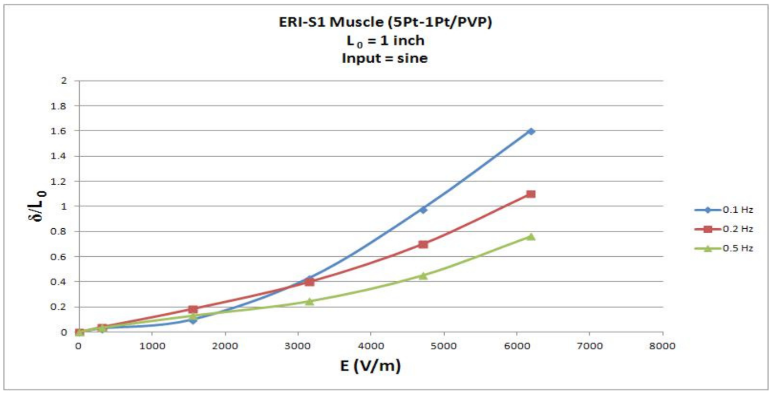


Figure 3: Non-dimensional IPMC cantilever (1cmx4cmx0.2mm) tip deflection versus the imposed sinusoidal electric field for three different frequencies (0.1-0.5 Hz).

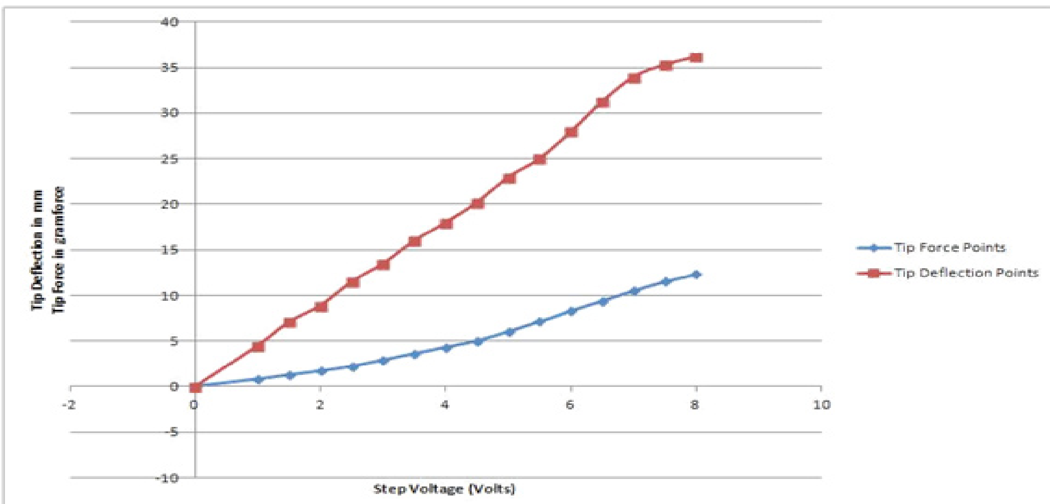


Figure 4: IPMC cantilever (1cmx4cmx0.2mm) tip deflection and tip blocking force versus the imposed step voltage.

(Figure-5a) depicts a scheme of the IPMC sample experiencing wave motion (flipping) and generates the corresponding electrical voltage signal. There are some studies on energy harvesting and related methods to analyze the energy harvesting of IPMCs from environmental dynamics (wind, turbulence, etc.) [6]. the majority of these studies so far establish generation of a few milli volts from small samples (1cmx4cmx0.2mm) [6]. (Figure-5b) depicts energy harvesting and sensing results in connection with an IPMC strip (7cm x 1cm x 0.3mm) experiencing arbitrary wind motions.

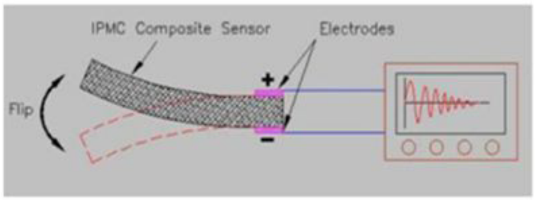


Figure 5(a): IPMC experimental configuration in a sensing/energy harvesting mode

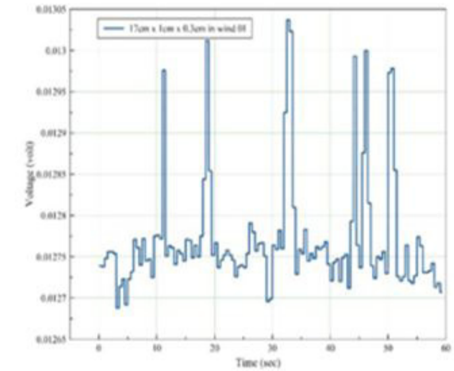
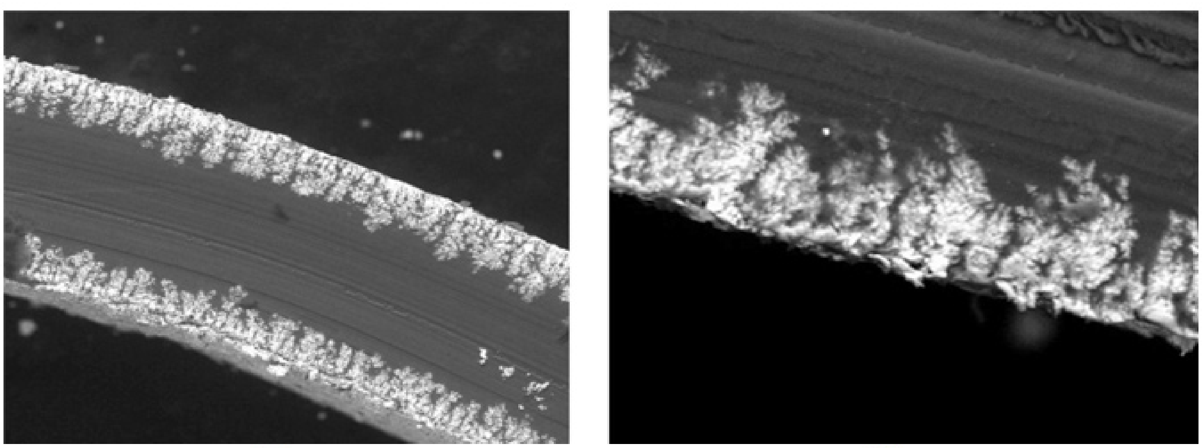


Figure 5(b): A (17cm x 1cm x 0.3mm) IPMC sample in a cantilever mode generating voltage signals under an arbitrary input disturbance

Note that IPMCs are essentially manufactured by a chemical REDOX operation [13]. In that operation the ionic polymer is initially surface roughened to increase its surface density for enhanced molecular diffusion of metallic salts into the molecular structure of the ionic polymer which is then place in a metallic salts solutions (for example, Tetra-amine platinum chloride hydrate) to be oxidized. The oxidized material is then placed in a reduction solution (Sodium borohydride) to reduce the metal (say platinum) on the macromolecules and around the 5 nanometer nanoclusters inside the ionic polymer [13] to generate Na^+ cations and further create fractal nanoclusters within the molecular network near surface boundaries, as shown in (Figure 6a and 6b).



(a) (b)

Figures 6(a-b): SEM picture of a typical IPMC thin strip showing the near boundary electrodes (a) and penetration of reduced metals in a fractal manner around nano clusters within the material (b).

The IPMCs are basically a two-phase system made up of a polar medium such as water. They contain ion cluster networks surrounded by an ion containing hydrophobic polytetrafluoroethylene (PTFE or Teflon). The structural stability of the ion containing polymer is provided by the PTFE backbones and the hydrophilic clusters which facilitate the transport of ions and hydrated water molecules attached to them in the ionic polymer. These nano clusters (3-5 nano meter) contain an interfacial region of hydrated, sulfonate-terminated perfluoroether side chains surrounding a central region of polar fluids with cations such as Na^+ or Li^+.

As reported by Shahinpoor and Kim [11], Shahinpoor, Kim and Mojarrad [12] and Kim and Shahinpoor [13, 14] ion containing polymers in a nano-composite with a conductor phase can be manufactured three-dimensionally to any complex shape. Essentially there are two ways to manufacture three-dimensional ionic polymeric nanocomposites with a conductor phase. One is to use a liquid form of the polyelectrolyte in an alcohol, such as liquid Nafion® in isopropyl alcohol. By meticulously evaporating the solvent (isopropyl alcohol) out

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of the solution, recast ionic polymer can be obtained. One can also use a precursor resin (XR Resin, DuPont) and hydrolyze it in KOH and then make it into a nanocomposite with a conductive phase.

Novel Family of Looped IPMCs as Soft Biomimetic Robotic Haptic Feedback Sensors

Looped IPMCs can be assembled in the form of a cantilever loop by bending a strip of IPMC around to reach itself end-to-end and form a looped haptic feedback sensor and looped actuator as shown in (Figures 7-10). The looped cantilever IPMCs are capable of bending and twisting actions as well as soft sensing. The sensing characteristics as a soft biomimetic haptic feedback sensor are shown to have a great potential for ubiquitous robot-human interactions (RHI) as well as providing haptic feedback for robotic surgery. (Figure 7) resembles a concept of a device or robotic end-effectors to detect the kinesthetic haptic feedback interaction and haptic forces with body organs.

Figure 7: Surgical robotic end effectors and its enlarged tip to measure kinesthetic haptic force feedback for hap tic interaction.

As reported by Shahinpoor and Kim [11], Shahinpoor, Kim and Mojarrad [12] and Kim and Shahinpoor [13-14] ion-containing polymers in a nanocomposite with a conductor phase can be manufactured three-dimensionally to any complex shape. Essentially there are two ways to manufacture three-dimensional ionic polymeric nanocomposites with a conductor phase. One is to use a liquid form of the polyelectrolyte in an alcohol, such as liquid Nafion® in isopropyl alcohol. By meticulously evaporating the solvent (isopropyl alcohol) out of the solution, recast ionic polymer can be obtained. One can also use a precursor resin (XR Resin, DuPont) and hydrolyze it in KOH and then make it into a nano-composite with a conductive phase.

A conceptual design for a looped haptic feedback sensor made with IPMCs can be similar the graphic shape in (Figure 8). The looped configurations IPMCs, (Figure 8), can interact with organs and tissues. They provide the relevant kinesthetic force signal based on the bending of IPMC strips or deformation of the loop. This signal can be correlated to the force exerted. It is expected that the looped cantilever IPMCs be capable of actuation and haptic feedback sensing simultaneously through proper design and configuration. The sensing characteristics as a soft biomimetic haptic as mentioned earlier is shown to have a great potential for ubiquitous Robot Human Interactions (RHI) as well as providing haptic feedback for robotic surgery.

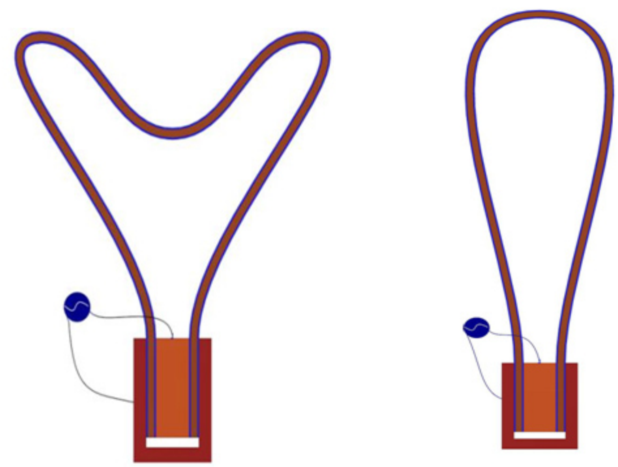
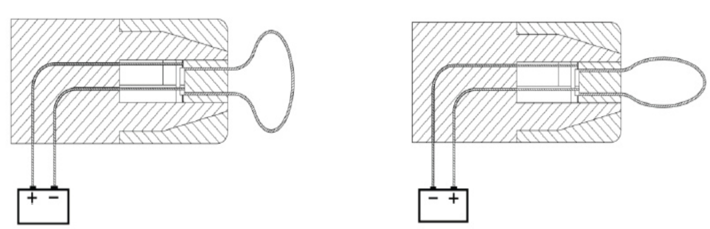


Figure 8: Configurations for looped IPMC haptic feedback sensors.

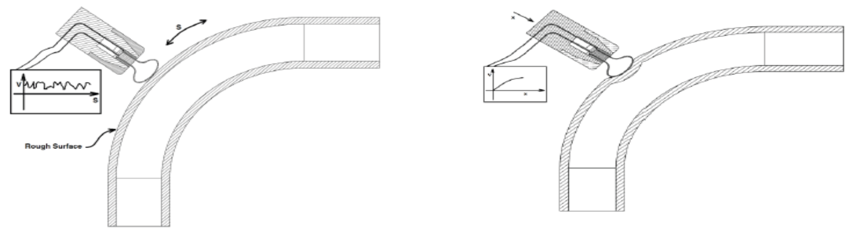
Furthermore, a looped IPMC haptic feedback sensors can enable a new technology in robotic surgery by providing surgeons efficient routines for kinesthetic and softness inquiry of organs and tissues during robotic surgery. This will be possible by simultaneous actuation and sensing on the fly by a small voltage (4-6 volts) for reconfiguration of looped IPMC feedback sensor in normal grasping and manipulation of bodily organs and tissues. On the other hand, a looped IPMC haptic feedback sensor can be applied as smart skin to a point of human -like dexterous and soft manipulation. Operationally they will be used either as haptic feedback sensors in robotic surgery or haptic feedback sensor in determining the softness of the materials or organ the surgeons are performing surgery. On the other hand, these soft looped IPMC sensors and actuators can be used in robotic automation and manufacturing in which soft and delicate parts need to be handled and looped IPMCs can be used as tactile sensors as well as haptic feedback sensors (Figures 9a, 9b and 10a, 10b).

(Figures 9a and 9b) depict various configurations of looped IPMCs providing haptic feedback sensing by contacting and pressing over various organs or bodies.



(a) (b)

Figures 9(a-b): Looped IPMCs with different (oblate or elongated tips) configurations.



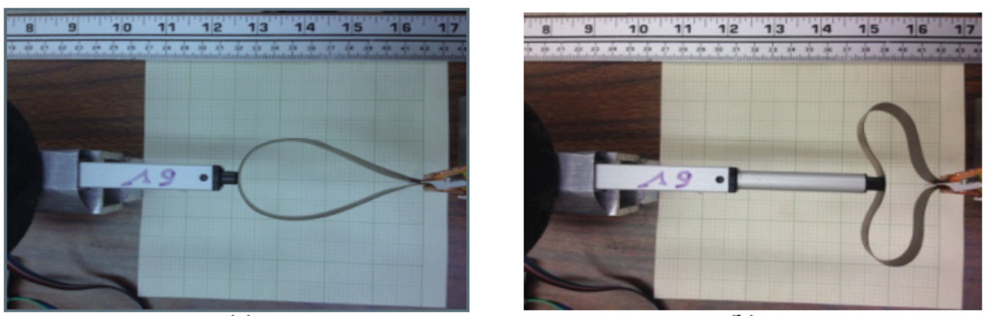
(a) (b)

Figures 10(a-b): Looped IPMCs (a) generating haptic feedback signal by pressing on and sliding over the surface of an external body (b).

Furthermore, these looped IPMCs can be experimentally subjected to various forms of external or internal loading as depicted in (Figures 11 and 12). In these test experiments, a single 14mm wide strip of a IPMC was bent to have a looped sensor with a length of 73mm. The IPMC loop was placed in front of a linear actuator. The linear actuator started the forward movement at t=10 Seconds and then returned at time t= 30Sec. In the next sections the results of several experimental results in connection with the proposed concept are reported. These tests include pushing the loop with the tip of a pointed linear actuator or a flat tip like a flat surface, while pushing the loop.

Part I-Frontal Tip Force Pushing the Looped IPMC Sensor

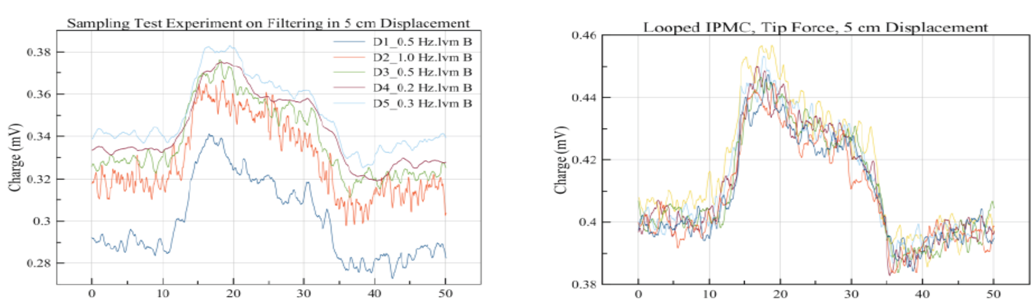
In this experiment, the actuator applies a force to move the looped sensor. Figures (11a and 11b) depict the experimental set up in which the looped haptic sensor is deformed in a direct manner and the shape of the looped sensor under full frontal deformation.



(a) (b)

Figures 11(a-b): IPMC and the linear actuator in surface contact (a), and in extended full 5cm deformation (b).

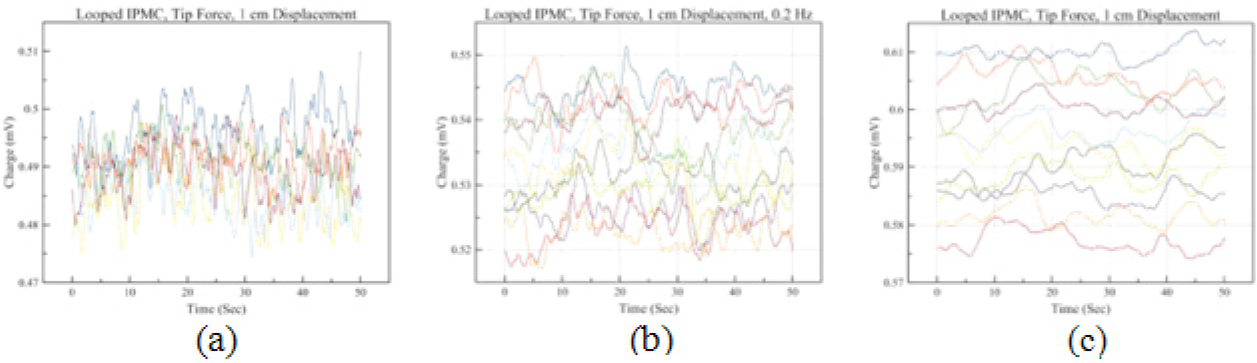
In this test experiment, a single 14mm wide strip of IPMC was bent to create a looped sensor with the length of 73mm. The IPMC loop was then placed in front of a linear actuator tip. For the test, Lab VIEW software with a data acquisition card and signal generator card were used. A pre-test was done to have a better understanding of the effect of noise filtering on the collected data. In this pre-test, different filtering were applied to a 5cm range of displacement during data collection (Figure 12(a)).



(a) (b)

Figure 12: IPMC sensing experimental results for a 5cm displacement under various filtering (a), and Test result of 10 experimental measurements for 5cm displacement (b).

Based on these results, (Figure 12(a)), 0.5Hz low pass filter considered to have a good sensitivity and smoothness for our data collection. For the next step, 10 tests were performed on each test range. After each test, the maximum displacement of the actuator was decreased by about10mm to observe the differences in the results. The results consistency and clarity for 4cm, 3cm, and 2cm displacements were more in the same form as what it is in (figure 12(b)).



Figures 13(a-c): IPMC sensing test results for 1cm displacement.

(Figure 13(a)) displays having a variation in charge level on the sample for a 1cm deformation. Charge variations for 1cm displacement are not as clear as they were for another level of deformations. To get better results other tests were performed by applying lower filtering frequencies. However, the results did not change very much, as shown in (Figures 13b and 13(c)), even though there are changes in the signal level in connection with the application of 0.5Hz and 0.2Hz. (Figure 14) display the variation of generated voltage vs. the applied displacement.

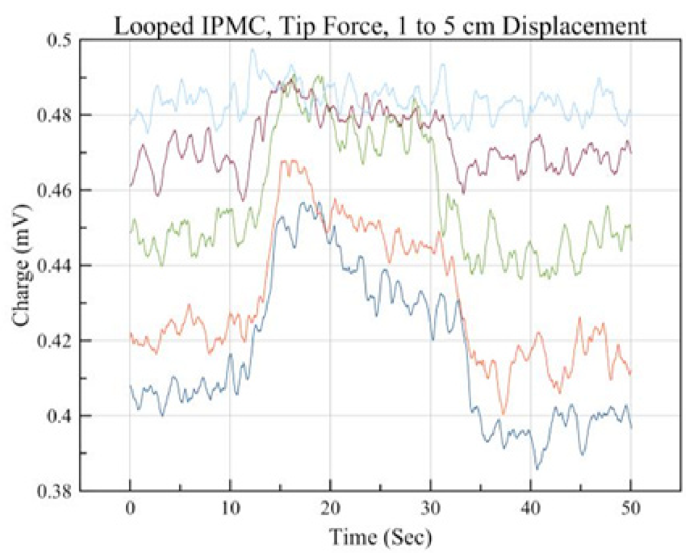


Figure 14: IPMC sensing test results, Variation of generated signal vs depth of frontal tip displacement.

Based on the above test results, depicted in (Figure 15), the consistency of received signal provides a good basis to consider and search for a linear relationship between the generated output signal and the applied displacement. The initial voltage variation and the deformation can be simplified in Table 1.

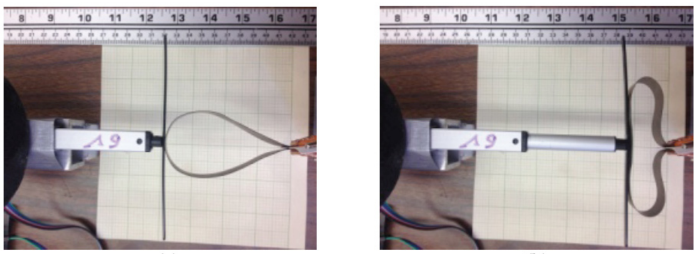
Deformation (cm)	5cm	4cm	3cm	2cm	1cm
Variation in charge level (mV)	5 mV	4mV	3mV	2mV	1mV

Table 1: Generated Voltage Vs Deformation or Tip Displacement.

The reported signals in Table 1, introduce an early attempt to get a version of response-deformation equation for this model of haptic feedback sensor.

Part II: Experimental Results: Frontal Flat Tip Pushing the IPMC Looped Sensor

In this experiment, a linear actuator with a flat tip or flat surface pushed the looped sample. The final deformation of the sample showed more bending than the test done by pointed tip pushing the looped IPMC and it is expected to generate larger signals.



(a) (b)

Figure 15: Looped IPMC being pushed by a Linear Actuator with A Flat Surface Tip touching (a) and Deformation of IPMC due to extended actuator by 5cm (b).

In this test for each test setup, 10 tests were done, and the results present a consistency for each configuration. (Figure 16) display variations of deformation vs the generated signal.

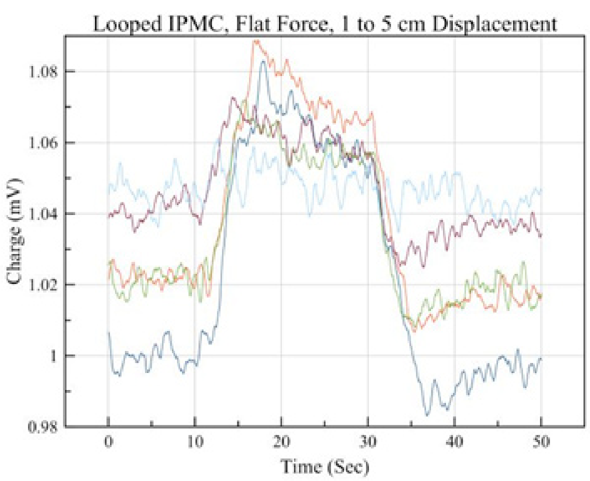


Figure 16: Variations of frontal displacement and time vs generated voltage signal.

The initial charge variations and the deformations for this test setup are summarized in the Table 2 below:

Deformation (cm)	5cm	4cm	3cm	2cm	1cm
Variation in voltage level (mV)	6.5 mV	6mV	5mV	3mV	1mV

Table 2: Generated Voltage Vs Deformation, Flat Front Displacement.

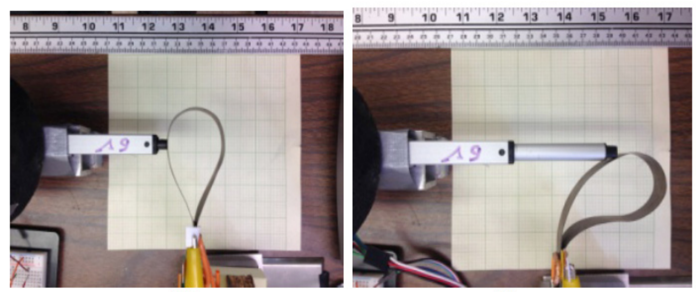
Based on the above test result the best polynomial fit to the voltage produced by the imposed displacement on the loop is given by:

$$V = -0.0003*D^2 + 0.003 * D - 0.002 \qquad (4)$$

Where, V is in Volts (v) and D is in meters (m).

Part III: Pushing the Looped IPMC Sensor with a Sideway Force

In this experiment the pointed tip of a linear actuator pushed the sample sideways. Here the 5cm deformation for the loop is the maximum possible deformation/interaction as if the interaction between the sensor and the actuator is considered to be the interaction between the sensor and a surface. (Figure 17(b)) demonstrates that the tip of the actuator will reach a configuration in which it can no longer deform the looped sensor.



(a) (b)

Figures 17(a-b): IPMC and the actuator in pushing the looped IPMC haptic sensor (a) and resulting deformation of looped IPMC due to actuator pushing by 5cm (b).

In this test for each test setup, 10 tests were performed and the results presented a consistency with previous experimental results for each configuration.

(Figure 18) displays the variations of deformation vs the generated voltage signal

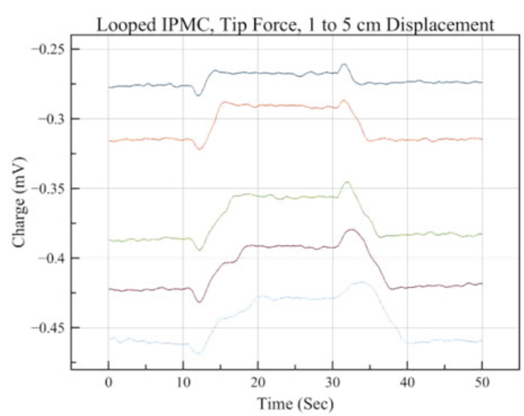


Figure 18: IPMC sensing test results, Variation of generated signal vs depth of frontal displacement.

The initial voltage variation and the deformation for this test setup simplified in table 3.

Deformation (cm)	5cm	4cm	3cm	2cm	1cm
Variation in voltage level (mV)	0.05mV	0.038mV	0.035mV	0.025mV	0.01mV

Table 3: Generated Voltage Vs Deformation, Flat Front Displacement.

For this case the best-fitted second order polynomial will be:

$$V = -0.9e - 6 * D^2 + 10e - 6\ e\ D - 3e - 6 \qquad (5)$$

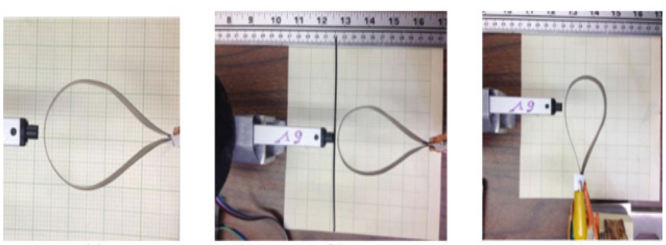
Where V is in Volts (v) and D is in meter (m).

Residual Plastic Deformations Observed in the Looped IPMCs

There were some residual plastic deformations observed in each looped IPMC sensor after the experimental procedures and 5 cm pushing actions.

(Figures 19a, 19b and 19c) depict the residual gap between the linear actuator tip and the looped IPMC sensors. These residual quasi-plastic deformations disappeared after a few minutes (7 minutes after maximum pushing distance of 5 cm).

We are currently looking into this recovery and will report the results later.



Figures 19(a-c): Final Gap after 10 times, 5 cm Deformation, Tip Force (a), Flat Force (b), Side Force (c).

Conclusions

Towards future soft robotics implementation, as well as realizing soft robotic haptic feedback sensors in surgical applications, various configurations of IPMC sensors and actuators were considered and experimentally tested. The sensing and haptic feedback signals generated by various looped IPMC sensors were reported. The new family of soft biomimetic robotic looped PMC sensors and actuators in the form of a cantilever loop produced by bending a strip of IPMC around end-to-end to form a looped haptic robotic feedback sensor and looped actuator. It is concluded that sensing characteristics as a soft biomimetic robotic haptic feedback sensor are shown to have a great potential for ubiquitous robot-human interactions (RHI) as well as providing haptic force feedback to surgeons during robotic surgery. Upon various types of deformation, they are shown to generate unique output voltage signal and transient current to be correlated to the actual haptic feedback force. Furthermore, IPMC haptic feedback sensors were shown to be capable of providing surgeons efficient routines for kinesthetic and softness inquiry of various organs and tissues during robotic surgery and yet they can be actuated simultaneously on the fly by a small voltage (4-6 volts) for reconfiguration of IPMC feedback sensor for normal grasping and manipulation of bodily organs and tissues or delicate parts in manufacturing automation. Finally, it is concluded that looped IPMC robotic haptic force feedback sensors can be considered as a smart skin.

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