

**FABRICATION, MODELING, ANALYSIS AND APPLICATION
OF SILVER- ELECTRODE IONIC POLYMER METAL
COMPOSITE ACTUATOR**

A Thesis Submitted in Partial Fulfillment of the Requirements
for the Degree of

DOCTOR OF PHILOSOPHY

by

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CERTIFICATE

It is certified that the work contained in the thesis entitled “**Fabrication, Modeling, Analysis and Application of Silver-electrode Ionic Polymer Metal Composite Actuator**” submitted by **Mr. Dillip Kumar Biswal** to the Indian Institute of Technology Guwahati for the award of the degree of Doctor of Philosophy has been carried out under our supervision in the Department of Mechanical Engineering, Indian Institute of Technology Guwahati. This work has not been submitted elsewhere for the award of any other degree or diploma.

The thesis, in our opinion, has reached the standard fulfilling the requirements for the award of degree of Doctor of Philosophy in accordance with the regulations of the Institute.

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Dedicated to.....

Mr. Duijabara Biswal

Mrs. Premashila Biswal

My parents for their blessing and guidance

Mrs. Swagatika Biswal

My wife for her great understanding and encouragement

Master Aryaman Biswal

My son for his love and affection

Mr. Dipak Biswal

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Abstract

Artificial muscle actuators mimicking the biological actuating mechanism are emerging as a new means of producing bio-actuator. Particularly, ionic polymer metal composite (IPMC) actuator that can be actuated or driven by employing a voltage which results in large strains to produce force and change in displacements have shown much interest to the research community. The ionic polymer metal composites originate from a class of advanced active materials, commonly known as electro-active polymers (EAPs).

An IPMC typically consists of a thin ion-exchange membrane such as Nafion or Flemion chemically plated on both surface with a noble metal such as gold or platinum as surface electrode. The water content of the base polymer of the IPMC serves as a medium for ion migration and thus influences the performance of the IPMC. Dehydration of moisture/water molecules through the porous electrode surface results in underperformance during working of the actuator. Further, for prolonged operation electrolysis occurs beyond certain input voltage again reduces the effectiveness of the IPMC. Though IPMCs are seemed to be very promising in various fields for actuation and sensing related applications, high manufacturing cost, nonlinear response, dehydration effect, and low force output during actuation have somewhat restricted its usage in commercial applications.

When an electric potential is applied across the thickness of IPMC, mobile cations combined with the surrounding water molecules move towards the oppositely charged electrode. The migration of mobile cations with the water molecules initiates swelling near the cathode electrode and shrinkage near the anode which leads to bending deformation towards anode.

The current state-of-the-art procedure to fabricate this artificial muscle is expensive as the metal electrodes are made of with platinum or gold. It is anticipated that changes in processing parameters and operating conditions may be able to

reduce reliance upon these metals. There is also serious limitation of arresting moisture contents which results in poor performance during experiment. It has been observed from the literature that, limited success has been achieved for successful fabrication of silver electrode IPMC to have superior or reliable performance compared to other electrode materials.

The proposed work deals with fabrication, characterization, theoretical and experimental analysis of Nafion based ionic polymer metal composite actuator of non-precious metal silver (Ag) as surface electrode. Chemical decomposition method is followed and several steps of trial-and-error experiments results in successful fabrication of Ag-IPMCs. The in-depth fabrication procedures are outlined in this work.

Scanning electron microscope is used to analyze the microscopic and morphological characteristics of IPMC samples which show dense packed silver (Ag) particles on the surface of the Nafion membrane. The crystal structure of IPMC is investigated by X-ray diffraction analysis. Micro-tensile test of the specimen is carried out to ascertain the stress-strain relationship of the base polymer Nafion and the fabricated IPMC. Thermo-gravimetric analysis (TGA) and differential scanning calorimetry (DSC) test are carried out and thermal stability of the actuator is assessed. Dynamic-mechanical analysis (DMA) and thermo-mechanical analysis (TMA) are performed to evaluate the viscoelastic properties and performance of the actuator.

An experimental setup is developed to carry out the bending and dynamic analysis of the developed IPMC actuator. Bending test is performed in fixed-free configuration for measuring tip displacement, tip force and surface resistance under DC potential. The developed pure Ag-IPMC is inexpensive, exhibits superior performance and can be used as actuator and artificial muscle materials.

Several studies have been carried out to find out the effect of water content on the actuation performance of the developed Ag-IPMC. It is well established fact that an IPMC actuator dehydrates continuously in open environment reduces the initial water content with applied electric potential. This significantly affects the actuation performance of IPMC actuator. In this work, efforts are made to develop a simple

but robust empirical model using Cobb-Douglas production method to predict the loss of water content in terms of input voltage and time in the open working environment. A term 'loss-factor' is introduced to represent the loss of moisture/water content from IPMC in terms of input voltage and actuation time. The effect of dehydration is assessed on tip displacement during bending of IPMC actuator for different input voltages.

During actuation it is observed that IPMCs experience low amplitude vibration while in bending motion. An IPMC actuator of silver electrode has been analyzed to demonstrate the effect of dehydration on vibration characteristics during actuation under DC input voltage. Multi mode approximation has been taken into consideration and extended Hamilton's principle is applied for developing the governing equation of motion of the actuator. Theoretical steady-state results are compared and validated with the experimental results. Both theoretical and experimental results show the gradual reduction of tip displacement owing to dehydration.

The precise predication of stable and unstable zone for an applied excitation voltage and frequency is of importance in many micromechanical systems utilizing IPMC as the actuator. A linear dynamic model thus will no longer be valid for IPMC actuator having low thickness and thus more flexible. Literature shows a little research has been carried out to find out the nonlinear vibration characteristics of IPMC actuator, excited by a small alternating electric potential. Though development of a comprehensive model is still going on, factors such as dynamic response and nonlinearities of IPMC have received less attention in the previous works. In the present work a nonlinear governing equation of motion has been developed which is solved by using method of multiple scales. A closed form expression for the response amplitude and detuning parameter has been developed which can be used to obtain the nonlinear frequency response curve for any IPMC actuator. The results obtained from perturbation analysis are compared with the experimental results and are found to be in good agreement. The developed simple expression can be used for finding the response of the system instead of solving the

temporal equation of motion which is time consuming and requires more memory space.

Active materials are nowadays utilized successfully for interventional diagnosis and therapy for their biocompatibility, suitability for smooth and safety operation. Active catheter is one such mechanism where active materials such as shape memory alloy (SMA), electro-active polymers (EAP) and piezoelectric materials are successfully used over the few years. In the present work, an active novel variable length catheter mechanism system has been developed using the fabricated IPMC actuator for *in-vivo* oral and body inspection or scanning. The catheter takes into account of the forward movement mechanism that makes it suitable to take scanning image covering more area with same input voltage. The low cost Ag-IPMC catheter system also potentially suitable in endoscopes for imaging of internal human tissues, blood vessel etc.

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Nomenclature

A	Area of cross section
C	Damping Co-efficient
C_D	Drag coefficient
C_s	Damping matrix
E	Elastic modulus
E'	Storage modulus
E''	Loss modulus
EI	Flexural rigidity
F	Force generated by the actuator
F_b	Bouncy force
F_{dg}	Drag force
F_n	Coefficient of the linear force term
h	Thickness
I	Cross-sectional moment of inertia of the beam
I_s	Mass matrix
K_s	Stiffness matrix
L	Lagrangian of the system
l	Length of the IPMC
$l(1-\gamma_v)$	Length of the characteristic pivot located from the fixed end
l_a	Active length
M	Bending moment of the actuator
$M(s)$	Bending moment of the link at a distance s from the fixed end along the length of the link
M_d	Bending moment of the actuator 'positive' dehydration

M_e	Bending moment in a fluid medium
M_h	Hydration mass
M_0	Bending moment of the actuator 'zero' dehydration
M_{wu}	Water uptake
m	Mass
n	Number of modes
p_x	X -Coordinates of the actuator
p_y	Y-Coordinates of the actuator
p_{xd}, p_{yd}	Tip position (X and Y-Coordinate) of the actuator for "positive" dehydration
p_{x0}, p_{y0}	Tip position of the actuator (X and Y-Coordinate) for 'zero' dehydration
p_{x1}, p_{y1}	Position vector of any point B on the first segment IPMC, used in equation 3. 32
p_{x2}, p_{y2}	Position vector of any point E on the second segment, used in equation 3.33
$q_i(t)$	Time modulation
R_c	Radius of curvature
r	Scaling factor
S_i	Path travelled by the tip of the actuator
S_{loss}	Solvent loss
s	Reference variable along the beam
T	Kinetic energy
T_c	Crystallation temperature
T_g	Glass transition temperature
T_m	Melting point temperature
t	Time
$\tan \delta$	Damping characteristics of the material obtained from DMA test
U	Potential energy
$u(\xi, t)$	Longitudinal displacement of the beam element $d\xi$

V	Input voltage
V_{dis}	Volume of displaced fluid
v	Velocity
W_c, W_a	Work done by IPMC actuator due to damping and bending moment
W_{fh}	Weight of the fully hydrated IPMC sample
W_{dry}	Weight of the completely dried IPMC sample
W_i	Weight of the IPMC before the voltage is applied
W_f	Weight measured after the voltage is applied for a certain time period
$W_{wet}(T)$	Weight of the IPMC samples at a certain temperature in wet condition
w	Transverse displacement
w_d	Width
$w(\xi, t)$	Transverse displacement of the beam element $d\xi$
X, Y	Newtonian Cartesian reference frame
α	Coefficient of thermal expansion
α_1	Coefficient of the non-linear geometric term (q^3)
α_2	Coefficient of the non-linear inertia term ($q^2\ddot{q}$)
α_3	Coefficient of the non-linear inertia term ($q^2\dot{q}$)
α_4	Coefficient of the non-linear inertia term ($q\dot{q}^2$)
δ	Logarithmic decrement
ε	Book keeping parameter
φ	Tip angle
φ_0	Tip angle 'zero' dehydration
φ_d	Tip angle 'positive' dehydration
φ_1, φ_2	Orientations of patch 1 and 2, used in equation 3.19
$\frac{d\varphi}{ds}$	Rate of change in angular deflection along the IPMC strip
γ	Characteristic radius factor

γ_v	Variable characteristic radius factor
γl	Variable characteristic radius
λ	Loss-factor
θ	Angle between the undeformed IPMC actuator (X-axis) and the tangent to the deformed actuator at zero slope
θ_v	Variable pseudo-rigid body angle
θ_{v1}	Pseudo-rigid body angles of the first segment IPMC
$\theta_{v1} + \theta_{v2}$	Pseudo-rigid body angles of the second segment IPMC
ρ	Density
ρ_f	Mass density of the fluid
ρ_c	Bending curvature
σ	Detuning parameter
τ	Non-dimensional time
ω_1	Fundamental frequency of the system
ω_i	Modal frequencies
$\psi_i(x)$	Admissible function
ζ	Damping ratio
ξ, η	Integration variables
K	Torsional spring constant
K_v	Variable torsional spring constant
Υ, a, b	Constants and depend on the material properties (used in equation 3.5)
Ω	Frequency of the applied electric potential

Introduction

Polymeric artificial muscle technologies are being developed that can produce similar strains and higher stresses using electrostatic forces, electrostriction, ion insertion, and molecular conformational changes. The performance of the emerging polymer actuators exceeds that of natural muscles in many respects, making them particularly attractive for use anywhere where a muscle-like response is desirable, including in medical devices, prosthesis, robotics, toys, biomimetic devices, and micro/nanoelectromechanical systems. Commercial application of these materials is at an early stage. Challenges remain with many of the technologies, most of which could be overcome via improvements in material properties. Artificial muscle actuators mimicking the biological actuating mechanisms are emerging as a new means of producing bio-actuators. Of particular interest, ionic polymer metal composite (IPMC) actuators that can be actuated or driven by employing a voltage which results in larger strains to produce forces and change in displacements.

The ionic polymer metal composites originate from a class of advanced smart materials, commonly known as electro-active polymers (EAPs). Electro-active polymers (EAPs) which have only been explored in the past decade are belonging to energy exchanging active materials and are very promising materials for many applications. These electro-active polymers (EAPs) respond to electric stimulation with a considerable deformation and produce a measurable electric potential by mechanical deformation, and can be used as artificial muscle materials [Sadeghipour *et al.*, (1992)]. IPMC type of artificial muscle employs a polyelectrolyte as an ion-conducting layer, in which polymer chains provide one ion type and mobile solvated ions the opposite type. The polyelectrolyte is sandwiched between two high-surface-

area flexible metallic electrodes, which interpenetrate the polyelectrolyte. The metal electrodes are typically composed of percolated platinum nanoparticles or gold.

As the present work aims to develop and study novel ionic polymer metal composite, the next section outlines the development of IPMC in a historical context, followed by the basic concept of actuation.

1.1 History of Ionic Polymers

Ionic polymer metal composites (IPMCs) have several configurations, but they are commonly categorized with respect to driving mechanism. Electric EAPs are driven by an electric field governs by coulomb forces, while ionic EAPs are driven by the mobility/diffusion of ions [Kim and Shahinpoor, (2003)]. Ionic polymer gels, ionic polymer metal composites (IPMCs), conductive polymers, carbon nanotubes, and electro-rheological fluids fall under the category of ionic EAPs [Bar-Cohen, (2001)]. EAPs have shown considerable bending actuation when subjected to a small applied electric potential of 1-3V and are also capable of sensing motion by producing a measurable electric potential when subjected to a mechanical deformation. These unique properties can be useful in a variety of applications that requires actuation or sensing.

One of the most widely used ionic polymers available today for fabrication of ionic polymer metal composite (IPMC) is DuPont™ Nafion® which is developed in the early 1960s with General Electric as a fuel cell membrane [Grot (2008)]. The material is a Teflon-based polymer with sulfonic acid side groups. When the material is soaked in water at 100°C for 1h, the membrane absorbs water as much as 38% of its dry weight, linearly expands as much as 15% of its original length and thickness increases up to 14% [DuPont Product Information sheet]. The chemical structure of the Nafion is shown in Figure 1.1 [Seen, (2001)]. Where m , n and X depend on the type of the polymer. For example, in Nafion 117 membrane, $m=1$, $n=6-7$, and $100 < X < 1000$. In fully hydrated condition, the ability to conduct cations across the thickness of the polymer membrane, gives them the electromechanical transduction capability [Newbury and Leo, (2002)].

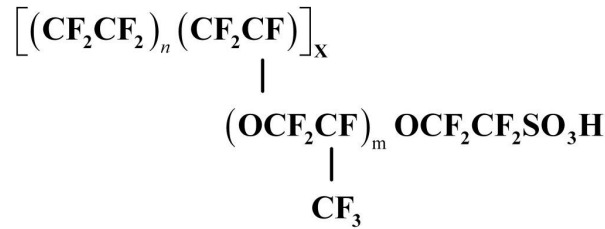


Figure 1.1 Structural formula of the Nafion polymer in acid form [Seen, (2001)].

Although, ionic polymers have been commercialized in the early 1990s, the extensive study for its actuation and sensing capabilities has been explored in the recent years. Oguro *et al.*, (1992), and Segalman *et al.*, (1992) demonstrated the use of Nafion based composite as an electromechanical actuator. Both of them showed that, by applying an electric potential to the material results in mechanical deformation. Sadeghipour *et al.*, (1992) also demonstrated the sensing capability of IPMC under mechanical deformation by developing an accelerometer. These results showed that when an external electric potential is applied across the thickness of the material it shows bending deformation; conversely a measurable voltage output can be obtained across its surface when it is deformed mechanically. These unique properties make ionic polymers as an alternative in the field of active materials and make them suitable for being used as both actuator and sensor.

Table 1.1 Comparisons between IPMC, SMA and EAC [Shahinpoor *et al.*, (1998)]

Property	Ionic polymer metal composite (IPMC)	Shape memory alloy (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	1-1.25 g/cc	5-6 g/cc	6-8 g/cc
Drive voltage	4-7V	NA	50-800V
Power consumption	watts (W)	watts (W)	watts (W)
Fracture toughness	Resilience, elastic	Elastic	Fragile

Table 1.1 shows the comparative study between the properties of IPMC with shape memory alloy (SMA) and electroactive ceramics (EAC) materials. It is observed that the weight to actuation capability of IPMC is much higher compared to both SMA and EAC materials. Further, response of IPMC material is quite fast compared to SMA materials. IPMCs can also be designed to imitate the action of biological muscles and also having unique characteristics such as low density, high toughness, large actuation strain and inherent vibration damping. Recently, due to the shortages and expensiveness of ion exchange polymer (IEP) such as, Nafion or Flemion, many polymer researchers are developing different types of new ion-exchange membrane, e.g., BPSH (sulfonate poly (arylene ether sulfone)), PATS (poly (arylene thioether sulfone)s), CP (chitosan/polyaniline), SSEBS (sulfonated poly (styrene-*b*-ethylene-*co*-butylene-*b*-styrene)) which are considered as the next generation materials as substitutes for Nafion and Flemion for fabrication of IPMCs.

1.2 Actuation Mechanism

Till date there is no universally accepted actuation mechanism available for IPMC and hence the fundamental principle that governs the actuation mechanism of an ionic polymer is still under research. The electromechanical characteristics in ionic polymers was discovered over a decade ago [De-Gennes *et al.*, (2000); Nemat-Nasser and Li, (2000); Tadokoro *et al.*, (2000)]. The electromechanical coupling effect and changes in the physical phenomena complicates the actuation mechanism.

An ionic polymer transducer consists of two basic components: a base polymer membrane (ionomer) and conducting metal electrodes as shown in Figure 1.2. The main distinction of an ionic polymer metal composite actuator over fuel cell is that the ionomer material (Nafion/Flemion) is sandwiched by conductive electrodes on both surfaces facilities to apply an electric potential that induces the bending deformation. The metal electrodes are mostly developed by depositing the electrode materials on the surface of the membrane chemically/mechanically, forming flexible electrode layers. Figure 1.2 shows the top and bottom surfaces represent the conducting electrode layers, where the outer surfaces are observed to be relatively

flat and smooth. The interfacing between the electrode layer and polymer is non-uniform due to the surface roughening and penetration of the electrode material into the surface of the membrane. The base material of the membrane is a perfluorinated ionomer that consists of chemically and thermally stable fluorocarbon backbone, hydrophilic anion domains (fixed anion), hydrophobic side chains (mobile cations), and solvent (water), as shown in Figure 1.3(a). The most common solvent used is deionized water, serves as a medium for ion migration within the base polymer. In neutral state the mobile cations are bonded with the fixed anionic groups, forming a cluster with the solvent (water) molecules, also easily exchangeable with the other cations [Lopez *et al.*, (1977)].

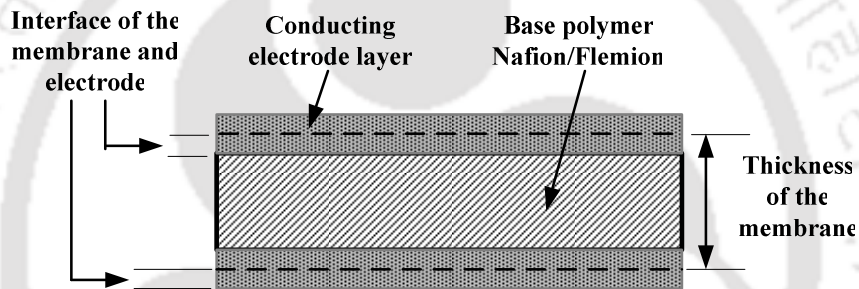


Figure 1.2 Schematic diagram of IPMC showing base polymer (Nafion/Flemion) sandwiched between the metal electrodes on both sides.

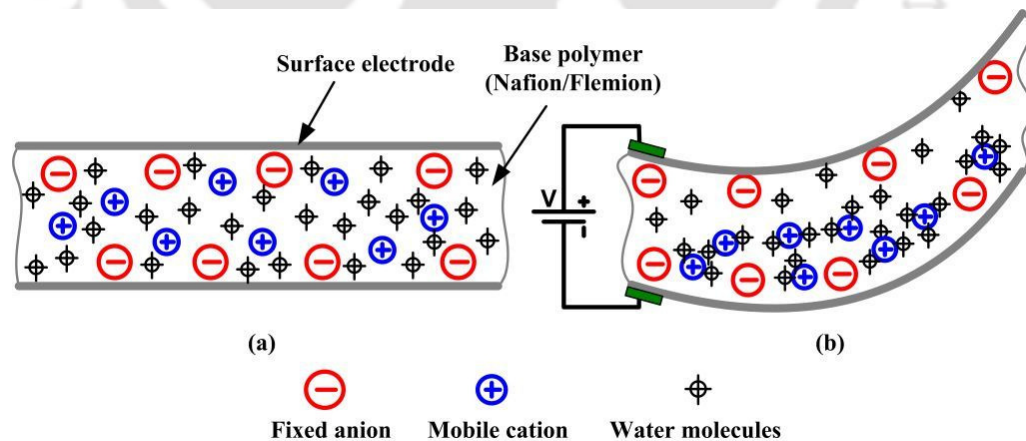


Figure 1.3 (a) Schematic diagram of the IPMC actuator (b) actuation mechanism under electric potential.

Some of the well accepted basic concepts about the actuation mechanism are reviewed in this section. When an electric potential is applied, the weak ionic bond

break apart as the positively charged cations are migrated towards the negatively charged cathode [Lakshminarayanaiah, (1969)]. De-Gennes *et al.*, (2000) proposed a model that postulates water pressure gradients (mobile cations drag water molecules with them as they move toward cathode, thus creating a pressure gradient) as the primary mechanism for actuation. Nemat-Nasser and Li, (2000) proposed a model which suggests that the ion transportation (redistribution of cations) causes bending of IPMC. When an electric potential is applied across the thickness of IPMC, mobile cations combined with the surrounding water molecules move towards the oppositely charged electrode. The migration of mobile cations initiates swelling near the cathode electrode and shrinkage near the anode which leads to bending deformation towards anode as shown in Figure 1.3(b). However, it could be possible that both the forces actually play a role in the actuation mechanism of IPMC actuator.

When a step voltage is applied across the thickness of IPMC actuator, at first it bends towards the anode and while the voltage remains constant, the actuator again relax back towards the cathode and attains a steady-state position. This characteristic indicates that there are two fundamental mechanisms that govern the actuation principle of the IPMC actuators. It is anticipated that, the migration of mobile cations towards the oppositely charged electrode is related to initial bending motion, while the relaxation is associated with the pressure gradient. This back relaxation phenomenon is observed in the IPMC actuator with water as polar solvent. Although back relaxation does not always occur when other variants of solvent are used in fabrication process, which is discussed later.

Eventhough, ionic polymer metal composites (IPMCs) have several outstanding characteristics to be used as actuator/sensor in various applications, there are certain drawbacks of Nafion which sometimes causes poor performance during actuation. Hydration level is one of the key issues as the solvent (water) which helps the mobile cations to move although evaporates in air during actuation. Thus, IPMC suffers from dehydration when it is operated in the open environment. It has been shown by Kothera *et al.*, (2003), that the dynamic response of the IPMC actuator can vary up to 20% in 10 min time interval while Bennett and Leo, (2003) showed

that the maximum deformation diminishes at around 10,000 cycles with a step input. Next section reviews some of the important topics such as fabrication techniques, mathematical models and application of IPMC proposed in the literature.

1.3 Review of Literatures

In this section, comprehensive reviews of literatures are done on ionic polymer metal composite taking into consideration about the fabrication techniques, modeling and applications.

1.3.1 Fabrication Techniques

Ionic polymer metal composites (IPMCs) composed of a thin ionomeric membrane such as Nafion-117 or Flemion plated both surfaces with a noble metal such as gold or platinum. The electrode layer serves as the conducting medium during actuation and also acts as sealing layer for moisture content. The process of depositing electrode layer over the ionomeric membrane is termed as electroding. The electroding techniques are broadly classified into two groups (i) chemical decomposition or electroless plating and (ii) mechanical plating method. Due to better adhesion of electrode layer to the base polymer, chemical decomposition method is mostly acceptable method. However, the method is proved to be time consuming and more expensive compared to mechanical method. Eventhough, a uniform electrode coating can be achieved following the mechanical plating method, they do not function properly in the hydrated state [Tiwari and Garcia, (2011)].

Chemical Decomposition Method:

In chemical decomposition/plating or electroless plating method a reducing agent is used for reduction of metal particles inside the polymer membrane. Earlier, chemical plating method was proposed by Takenaka *et al.*, (1982) for plating Nafion. In this method, the membrane was clamped between two chambers of a reactor. One surface of the membrane is in contact with the metal (Pt) salt solution and other side of the surface is in contact with the reducing agent (NaBH_4) solution. The reducing

agent solution continuously diffuses through the membrane from one side and make in contact with the metal salt solution and initiates reduction process. Noble metals and alloys were plated to both surfaces of the membrane by chemical reactions of metal salt solution and using a reducing agent without any binder. This method is the most efficient electroless plating technique employed for fabricating IPMCs. Another method similar to the previous one was proposed by Millet *et al.*, (1989, 1995) for plating platinum onto a Nafion membrane. In this process, the membrane is immersed into the metal solution and the metal ion is penetrated into the Nafion membrane and later reduced it into metal particles using a reducing agent. This method is comparatively better than the previous method as it can form long-time stable electrodes easily.

Oguro *et al.*, (1999) proposed a method to fabricate IPMC with gold as the surface electrode, where an ion-exchange counter ion such as Na^+ with a cationic (gold complex) solution is treated by a reduction process using reducing agents such as sodium sulphate. This ion exchange and reduction process can be repeated several times until a suitable thickness of the surface electrode is plated on the surface of the membrane.

Almost all the previous efforts to fabricate IPMCs are used platinum or gold as the surface electrode which contributes to the high cost of IPMCs limiting its potential applications in various fields. To reduce the cost of fabrication of IPMCs, Bennett and Leo, (2003) developed a co-reduction process for fabricating IPMCs with combination of both precious and non-precious metal electrodes. IPMC actuator fabricated by co-reducing copper and platinum (Cu-Pt) inside a Nafion membrane achieved a life cycle of 250,000 cycles (1.25V, 1.0 Hz sine wave) in deionised water. To reduce production cost, non-precious metal such as silver was used by Chen *et al.*, (2009) for fabrication of IPMC and the performance of the material was assessed by investigating deformation angle, surface resistance experimentally.

Mechanical Plating Method:

Although, fabrication of IPMCs using chemical decomposition method produces mechanically stable electrodes, they are complex, expensive, and time consuming. Further, effective fabrication requires a brief knowledge about the right/exact amount of chemical composition, and reaction between the chemicals and the membrane. Hence, sometimes mechanical plating techniques are also preferred over chemical plating method to fabricate IPMCs. Mainly three techniques are widely accepted for plating surface electrode over the ion exchange polymer membrane i.e., physical vapour deposition, solution casting, and direct assembly process.

Physical vapour deposition method, especially sputtering or evaporation is most commonly used technique for depositing surface electrode materials over the membrane. Sputter coating can be used independently or can be used in combination with other techniques like electroplating, electroless deposition and plasma etching. Zhou *et al.*, (2001) fabricate an IPMC by first putting a seed layer of gold ($0.4\text{ }\mu\text{m}$) by E-beam evaporation and followed by electroplating a $2\text{ }\mu\text{m}$ thick gold layer. The IPMC fabricated by this method can withstand a high voltage of the order of 20V without peeling of the electrodes. Siripong *et al.*, (2006) used sputter coating technique for depositing gold as a surface electrode on Nafion surface. The process proved to be fast and cost effective for fabrication of IPMCs.

Sputter coating provides an easy means for deposition of electrode materials over the membrane with less time consumption compared to the chemical decomposition method. Both precious metals (gold, platinum) and non-precious metals (nickel, silver, copper) can be deposited as the surface electrode without much change in the deposition technique. The thickness and surface roughness of the deposited electrode layer can be easily controlled. Lee *et al.*, (2006a) observed that improvement in performance in terms of tip displacement and force output can be achieved for IPMC fabricated by chemical decomposition method and followed by sputter coating of gold over it. However, cracks are formed on the surface of the IPMC as the sputter coating is performed in dry state.

The actuation force in an IPMC actuator is directly proportional to its dimensions and the applied electric potential. Eventhough, a thick IPMC can

develop a large tip force and can operate for a longer time, the bending tip deflection reduces considerably. Fabricating a thicker IPMC, solution casting and hot pressing techniques are generally followed. However, different shapes and sizes of IPMC transducers can be fabricated using solution casting methods.

Kim and Shahinpoor, (2002) successfully developed IPMCs having thickness greater than 2 mm from liquid Nafion with platinum as the surface electrode following the solution casting method. Further, Nafion solution can be mixed with the ion conducting powders like platinum, gold, palladium, copper, silver, carbon etc., for various electroding purposes. The ion conducting powder-coated electrode is cured under elevated temperature for achieving stability.

Chung *et al.*, (2006) followed solution casting method to fabricate IPMC with silver nano-powders. The silver nano-powder membrane and the liquid Nafion membrane are prepared and embossed together and baking of the membrane is done with a two-step temperature of 60°C and 100°C. To reduce the surface resistance, a silver layer is deposited on the casted IPMC by chemical decomposition method. Up to 15 μm thicknesses of silver electrode was achieved by following the chemical decomposition method. This IPMC shows large deformation of bending curvature angle (more than 90°) with an input voltage of 3V.

Similar to physical vapour deposition method, surface electroding using solution casting suffers poor bonding between the polymer membrane and electrode. Also, this method requires accurate adjustment of the process variables like temperature and concentration of the solvents. Hence, reproducibility may not be achieved exactly. To overcome this drawback hot pressing technique may be employed to fabricate IPMCs. In hot-pressing technique, a hot-pressing system is used to integrate several thin ion exchange membrane of Nafion together. This method is expected to enhance the blocking force as the bending stiffness of the base polymer increases. The method is also proved to be efficient as far as repeatability is concerned.

Lee *et al.*, (2006b) successfully fabricated thick film IPMC actuator by hot-pressing several thin film of Nafion membrane and achieved thickness between 0.54 to 0.9 mm. Several cycles of the Pt electrode plating was applied following chemical

decomposition method to improve the actuation performance of IPMC actuator. Tiwari and Kim, (2010) fabricated a disc shaped IPMC using Nafion granules following hot-pressing technique coated with conductive silver paint.

Direct Assembly Process (DAP) for fabrication of IPMC with novel polymeric design has been conceived by Akle, (2005). The method utilizes conductive nano particles mixed with Nafion solution is directly deposited on the Nafion membrane. A wide range of conductive nano particles such as single walled carbon nanotubes (SWNT), polyaniline (PANI), RuO_2 , carbon black etc., can be used to fabricate IPMCs. The technique simplifies the fabrication process and enhances the manufacturing repeatability of the IPMC actuator. Liu *et al.*, (2009) fabricated IPMC bimorph actuator following layer-by-layer (LbL) self-assembly process by using conductor network composite (CNC) electrode. In this method, CNC electrode can be deposited at submicron thickness ($\sim 0.4\mu\text{m}$) level with high precision and quality. IPMCs fabricated following this method show large bending actuation with fast response time.

Nanofibrous mat-based on a copolymer of polytetrafluoroethylene and fluorinated vinyl ether sulfonate (Nafion) has been proposed by Nah *et al.*, (2008). Electro-spinning process has been used to fabricate the Nafion nanofibrous mats. The obtained Nafion mat consists of randomly oriented submicron fibres with an average diameter of 200 nm. Electrodes are coated on to the mats surface following direct assembly process (DAP). The fibrous mat based IPMC actuators were observed to have faster response time (around 52%) compared to the film-based actuators.

Each fabrication processes i.e., chemical and mechanical plating methods have its own advantages and disadvantages. The fact attributes that the choice of a particular method or combination of different method to fabricate IPMCs mainly depends on the stability and quality of the electrode, cost and time consumed for fabrication.

1.3.2 Improvements on Fabrication

This section covers some of the research outcomes that enhance the actuation characteristics of ionic polymer actuators. Mainly the actuation characteristic depends on the type of electrode, mobile cations, solvent type, and ionomers. Effect of each parameter on the performance of IPMC actuator is also discussed.

1.3.2.1 Effect of Electrode

Fabrication procedure, various parameters and electrode plays an important role on the performance of an IPMC actuator. The electrode materials must satisfy two important criteria i.e., stability under operating condition, and at the same time it should be resistive to corrosion. In general, most of the cases IPMC actuators are electroded with precious materials such as gold or platinum due to their stability, resistance to corrosion and high conductivity properties. Shahinpoor and Kim, (2000a) first experimented with silver and copper by electroding a thin layer over a platinum electrode which reduced the surface-electrode resistance significantly. Experimentally, it is conformed that the reduction in surface resistance significantly improves the actuation performance as the force output increased by up to 20%. Shahinpoor and Kim, (2002a) improved actuation capability of an IPMC by physically loading a conducting polymer powder into the ion-exchange polymer and then a second layer particles by chemical coating over first layer. Bennett and Leo, (2003) develop a co-reduction process employing non-precious metal electrodes to fabricate inexpensive IPMC. An alloy of platinum and copper is deposited through reduction process while a thin layer of gold is used additionally to increase the surface conductivity and to prevent copper oxidation of the electrode.

Chung *et al.*, (2006) fabricated IPMC actuator using silver nano-particles (35nm) with diluted Nafion solution following micro-fabrication techniques and showed large deformation of bending curvature angle (more than 90°) with an input voltage of 3V is achievable. Kim *et al.*, (2006) coated a waterproof material over the surface of the IPMC to suppress the water evaporation for improving the lifetime of the actuator. Akle *et al.*, (2006) incorporated Ruthenium Oxide powder (RuO₂) for electroding, and the IPMC actuators show an increase in the strain response by a

factor of four. Carbon nanotubes (CNT) have also been used as the surface electrode [Lee *et al.*, (2006); Akle and Leo, (2008)]. Kim *et al.*, (2007) reduces the resistance of the electrode by 20% and improves the actuation performance by 60% and the lifetime by 90% by changing the surface morphology treating with O₂ plasma. Nguyen *et al.*, (2007) improves the blocking force and actuation performance of an IPMC actuator by incorporating nanoparticulates (3-10 wt %) as layered silicate or fumed silica in the Nafion matrix. Kim and Kim, (2008) observed improvement in the actuation performance by depositing a palladium layer over Nafion surface before plating the platinum electrode by chemical reduction method. The Pt/Pd IPMCs show a better actuation displacement and higher blocking force compared to Pt IPMCs. Gold foils (24-carat) have also been used for electroding by Palmre *et al.*, (2009).

Sometimes due to low cost and easy availability and large surface area, carbon black has also been used as electrode material [Akle, (2005); Park *et al.*, (2008); Palmre *et al.*, (2009)]. However, the adhesion between ionomer and carbon black is not impressive in hot pressing technique. Lian *et al.*, (2010) developed an IPMC actuator based on GO-Nafion nanocomposites and CNT as the electrode plated by dip-coating method. The developed actuator exhibits better actuation capability (around 2 times) compared to Nafion based actuator. However, poor dispersing capacity in the polymer matrix and high price, limits the use of CNT as the electrode. Saher *et al.*, (2010) also worked on improving the actuation performance of a platinum based IPMC by Cl₂ and SF₆ plasma treatment. The plasma treatment enhances surface micro-features and the electric capacitance although reduces the surface resistance considerably. Experimental investigations show these changes significantly enhanced the actuation performance and operational life of the IPMC.

New concepts for improvement of actuation capability of IPMCs also reported in the literatures. Connecting multilayer transducers parallelly, the force output can be increased [Akle and Leo, (2004)]. Jeon and Oh, (2009) developed a selective growth technique of platinum electrodes to improve the mechanical properties of a multi degree-of-freedom (MDOF) IPMC actuator for biomimetic locomotion. Chen and Tan, (2010) presented a new method for batch-fabrication of monolithic MDOF

IPMC actuators. To enhance the charge transport phenomenon under bending load, a thin film of Ag (60 % by weight) in polystyrene resin with a conductive hardener were coated on both side of Nafion-117 in the fabrication of IPMC by Anand *et al.*, (2010). Chen *et al.*, (2011) developed a synthetic technique to fabricate IPMC membrane which can produce 3D kinematic motion. Power consumption by the actuator are minimised by assembling IPMC actuators with a polydimethylsiloxane (PDMS) elastomer membrane. Feng and Tasi, (2011) developed a new 3D4E IPMC actuator by enclosing a section of optical fiber in the fabrication process, which improves the mechanical energy.

1.3.2.2 Effect of Counter-ion

Several experiments were performed using a variety of cations such as Na^+ , Li^+ by Shahinpoor and Kim, (2000b) to assess the improvement of the actuation capability of IPMC. This work revealed that actuators containing lithium ions (Li^+), possessed 40% higher efficiency (force output/power input) than the actuator containing sodium ions (Na^+). Further, it is also observed that cations having higher hydration numbers contributing more force and displacement of IPMC. Asaka *et al.*, (2001) studied the state of water and ion conductivity of Nafion® 117 and Flemion® membranes with variety of cations and discuss the factors affecting the deformation of solid polymer electrolyte membrane (SPM). The various types of cations were used such as alkali metals (Li, Na, K, Cs), alkaline earth metals (Ca, Mg, Ba), and alkylammonium ions like tetramethylammonium (TMA), tetraethylammonium (TEA), tetrapropylammonium (TPrA) and tetrabutylammonium (TBA). A similar study was carried out by Nemat-Nasser and Wu, (2003) and they concluded that for the same size of IPMC strip, the actuation speed varies for different cations. Flemion-based IPMC with TBA form show the highest tip displacement without any back relaxation effect.

1.3.2.3 Effect of Solvent

Past researches established the fact that the actuation capability of IPMC is strongly dependent on the hydration level. As the water continuously dehydrates in open

environment (evaporates through the surface) and electrolysis occurs beyond 1.23V, results in reduction of actuation capability of IPMCs.

To overcome the above mentioned disadvantages highly stable ionic liquids are explored nowadays to replace water as solvent. Ionic liquids offer greater electrochemical stability than water and they are non-volatile. Lu *et al.*, (2002) first conducted experiment using ionic liquids composed of 1-butyl-3-methylimidazolium cations together with anions such as tetrafluoroborate or hexafluorophosphate. The study shows some exciting results with negligible loss of solvent from the actuator. Nemat-Nasser, (2002) demonstrated the suitability of ethylene glycol as a solvent in IPMCs. These IPMCs considerably reduce the dehydration problem though affect the actuation performance significantly. Poly (ethylene oxide) and poly (ethylene glycol) were also used as solvents by Shahinpoor and Kim, (2002b). Poly (ethylene glycol) of low molecular weight also serves as the solvent. A parametric study of the electromechanical response of IPMCs using a wide variety of solvents including glycerol and butyl acetate was experimentally investigated by Farinholt and Leo, (2004).

Bennett and Leo, (2004) uses five different ionic liquids in his experiment and concluded that viscosity of the ionic liquids and the degree of swelling are important parameters to be considered. It was observed that out of these five ionic liquids 1-ethyl-3-methylimidazolium trifluoromethanesulfonate demonstrated the best results. Ionic liquid (1-ethyl-3-methylimidazolium trifluoromethanesulfonate) has been used by Bennett and Leo, (2005) as the solvent in IPMCs. Experimental results showed that IPMCs with ionic liquids are more stable compared to water based IPMCs when operated in air. The primary drawback of this ionic liquid is that the response speed diminishes compared to water. This investigation demonstrates the potential use of ionic liquids as solvents. Lee and Yoo, (2009) uses four different ionic liquids i.e., [EMIm][Br], [EMIm][NO₃], [EMIm][AcO], and [EMIm][TA] as the solvent of IPMC and observed that performance of the IPMC depends on the size and ion mobility in the ionic liquids.

1.3.2.4 Effect of Ionomer

Ion conductivity and selective ion diffusivity are driving factors for selecting an ion-exchange polymer (IEP) i.e., ionomer in the fabrication of IPMCs. Two types of ionomers (i) Nafion[®] produced by DuPont and (ii) Flemion[®] by Asahi Glass are commonly used as the base polymer for fabrication of IPMCs. Flemion-based IPMC actuator exhibits better ion-exchange capability, better surface conductivity, higher water uptake capacity, and higher stiffness. They also demonstrate superior bending actuation capability for same input voltage compared to Nafion IPMCs [Nemat-Nasser and Wu, (2003)]. Experiments showed that Nafion as the ionomer can display a back relaxation in the opposite direction of the initial response motion. On the other hand, actuators made of Flemion ionomer demonstrated a relaxation in the same direction with respect to the initial motion [Asaka *et al.*, (2001); Nemat-Nasser and Wu, (2003)]. However, factors like commercial availability, synthetic convenience and chemical stability, Nafion has been mostly employed as the base polymer in the fabrication of IPMCs [Duncan *et al.*, (2008)]. Even-though IPMCs of Nafion show good actuation capability, there are certain disadvantages such as (i) improcessible in molten state, (ii) back relaxation effects (iii) melting temperature (iv) high cost restricts uses of Nafion for wide applications. To overcome these disadvantages many researchers are trying to find alternative ionomers which can give better performance with low manufacturing cost.

Akle *et al.*, (2003a) developed a novel class of polymer called BPSH (sulfonate poly (arylene ether sulfone)s and PATS poly(arylene thioether sulfone)s. These polymers have been synthesized from direct polymerization of sulfonated monomers. Both the polymers differ from Nafion in terms of ion concentration, stiffness, and water content. Experimental investigations demonstrate that BPSH-35 shows improved performance in terms of larger strain per unit, large force generation, and larger band width compared to Nafion. A further experimental study is carried out with these ionomer materials showed that there is an approximately linear relationship between strain response and capacitance of the transducer [Akle *et al.*, (2005)].

Kim *et al.*, (2005a) used chitosan/polyaniline (CP) ion-exchange polymer membrane to fabricate IPMCs. Phillips and Moore, (2005) fabricated IPMCs introducing sulfonated ethylene vinyl alcohol copolymers as the base polymer. Experimental results show that the behaviour of the developed ionomer is similar to Nafion, although response is slower compared to Nafion. Wang *et al.*, (2007) developed an electro-active polymer through electroless plating technique, based on SSEBS (sulfonated poly (styrene-*b*-ethylene-*co*-butylene-*b*-styrene)) ionomer. The SSEBS is inexpensive and exhibits moderate back relaxation compared to Nafion and thus is a promising ionomer in the fabrication of IPMCs.

1.3.3 Modeling Technique for IPMC Actuator

Mathematical models of IPMC actuator can be classified into three major groups i.e., black box model, physics based model, and gray box model. Black box model describes IPMC behavior through a set of functional relationships among selected representative macroscopic IPMC variables like applied mechanical force, displacement at a given location, input voltage across the electrode thickness, and the storage charge. In this model the values are obtained from the curve fitting of experimental data, and they are specific with respect to the shape and size of the experimentally characterized IPMC. Physics based models describe the IPMC behavior through the fundamental principles of electrostatics and mechanics. Gray box model describes IPMC actuator as a continuous piezoelectric bender, whose mechanical and electromechanical properties are obtained from experiment. Charge redistribution and solvent motion are not taken into consideration in this model.

Shahinpoor, (1993) proposed the theory of nonhomogeneous large deformation of ionic polymeric gels in the presence of pH and electric fields. Based on experimental results, two distinct mechanisms were identified for the large bending deformation of the ionic polymeric gel strips. One is that the presence of pH field, anions and cations distribute within the gel network creating a spatial distribution of the ions. While for the second, electric potential across the thickness of the ionic polymer strip causes the actuator bends towards anode which mostly depends on the initial spatial distribution of the ions. Further Shahinpoor, (1994) proposed a

microelectro-mechanical model, which relies on the concept of the internal electric charge redistribution of fixed and mobile ions due to the presence of the electric field.

Kanno *et al.*, (1994) presented an empirical model and the characteristics of the ICPF (ion-conducting polymer film) actuator. Kanno *et al.*, (1996a) also proposed a 2-dimensional linear black-box model for the ICPF actuator. The dynamic model consists of three stages i.e., electric stage, stress generation and mechanical stage, considered to be the driving factors responsible for the bending of the actuator. Further, Kanno *et al.*, (1996b) extended the previous work and came up with a 3-dimensional model. In both the cases, the mechanical part was modelled using a finite element technique and the results were verified experimentally.

Mojarrad and Shahinpoor, (1997) investigated the sensing capability of a platinum electroded IPMC actuator by deforming the tip of the sample. The sensor films were characterised as a function of the tip displacement. Further, hysteresis was observed between voltage output and displacement for a complete cycle of bending. Vibration characteristics of platinum electroded ionic metal composites were investigated by Shahinpoor *et al.*, (1997). It was observed that with the increase in applied frequency, the amplitude of vibration increases till it reaches to resonant frequency beyond which the deformation gradually diminishes. It is also observed that the actuation performance of IPMC depends on amplitude and frequency of the applied voltage and dehydration.

Shahinpoor, (1999) developed a mathematical model similar to classical Euler-Bernoulli beam theory model, by including a non-homogeneous distributed electrically-induced moment. Analytical results were verified for Cationic Polymer-Platinum Composite (CPPC) and found to be in good agreement with the experimental results. De-Gennes *et al.*, (2000) proposed a linear irreversible thermodynamics model and studied the effects of driving forces (electrostatic force, water pressure gradient) and fluxes (applied electric current, water current) on the actuation behavior of the ionic gels.

Tadokoro *et al.*, (2000) proposed a white-box model for Nafion-Pt composite actuator. The model consists of two stages: (i) ion migration model, where ion

motion set by the electric potential and water motion by ion-drag has been considered. (ii) Stress generation model: where the swelling and contraction of membrane, momentum effect, electrostatic force and conformation change of the membrane is discussed. Simulation results have been validated through experimental results. They observed that tip displacement exhibits a nonlinear relationship with input voltage.

Nemat-Nasser and Li, (2000) developed a micromechanical model considering the coupled ion transport, electric field, and elastic deformation to predict the response of the platinum electroded IPMC. It is assumed that the ionic groups are fixed to the fluorocarbon backbone while counter ions are free to move with water. The model shows that the electrostatic forces, input voltage, redistribution of ionic groups, and redistribution of water molecules, are the main factors which are responsible for the bending actuation of IPMC actuator. A simulink model was developed by Alvarez and Shahinpoor, (2002) and demonstrated the static deflection for an elastic cantilever beam using nonlinear equation for large angle deflection.

Newbury and Leo, (2002) modified the model proposed by Kanno *et al.*, (1996a, b) which did not take into consideration of the force output. Newbury and Leo, (2002) proposed a linear electromechanical model which fit a simple linear relationship between the blocked force and displacement of the IPMC actuator. However, the model does not take into account of the complex dynamic relationship. Further, Newbury and Leo, (2003a, b), enhanced the previous model to account for polymer viscoelastic behavior, frequency dependent electric permittivity, and frequency dependent electromechanical coupling. In this model an equivalent RC circuit and Golla-Hughes-McTavish method is used to describe the viscoelastic behavior of IPMC. Yamakita *et al.*, (2003) developed a model following Hammerstein technique to identify nonlinearity in IPMC materials used in a walking robot. Jung *et al.*, (2003) developed an equivalent electrical circuit model from the experimental data to study the effect of different driving input waveforms (square, triangular and harmonic) and frequencies on the characteristics of the IPMC actuator. Yagasaki and Tamagawa, (2004) adopted an equivalent beam model to

investigate the viscoelastic behavior of IPMCs by fitting the experimental data to an exponential function.

Yim *et al.*, (2005) proposed an analytical model for establishing the relationship between input voltage and output of bending moment using a lumped RC model giving a first order response. Paquette *et al.*, (2005) developed a more sophisticated RC model to predict the bending behavior of a multi-layer IPMC strip with surface electrodes of nonlinear characteristics. An experimental investigation was carried out by Kothera and Leo, (2005a) to characterize the nonlinearity seen in the actuation response of cantilever ionic polymer benders. Volterra series are employed for system identification to aid in the characterization of the harmonic distortion. Ionic polymers employing different solvents [water, ionic liquids (EMI-Tf, BMPyr-tris)] are used for comparison. They found that the nonlinearity not only depends upon the types of solvents, but also frequency dependent. Kothera and Leo, (2005b) presented non-linear distortion model using the Volterra series and studied the closed-loop bandwidth of the cantilever actuator in both clamped-free and clamped-clamped boundary condition. Shahinpoor and Kim, (2001) show the nonlinearities by segregating IPMC actuator behavior with small hysteresis. Lee *et al.*, (2005a) developed an equivalent beam model and an equivalent bimorph beam model for IPMC actuation and obtained the physical properties along with finite element analysis (FEA) is used to predict the force-displacement relationship for an IPMC actuator.

The model proposed by Nemat-Nasser and Wu, (2006) shows that the composition and microstructure/morphology of the ionomer and electrode, hydration level, neutralizing cations, time duration of the applied electric potential are the main factors which affect the response of the IPMC. They showed when a small actuating potential is applied to a fully hydrated IPMC actuator, it shows a considerable bending vibration at that applied frequency. Branco and Dente, (2006) proposed a continuum electromechanical model to predict the electrical and mechanical behavior of the IPMC actuator. An equivalent circuit model for the IPMC is derived using continuum model to find out the relation between voltage and

current. The model assumes that the electrostatic force causes the actuation of the IPMC.

Bonomo *et al.*, (2006) proposed a nonlinear model based on an equivalent lumped parameter circuit, which describes the electrical behavior of an IPMC actuator. The model is validated through experimental results. Further, Bonomo *et al.*, (2007) presented a nonlinear electromechanical model to estimate the relevant phenomena that characterises IPMC actuator. The model consists of two stages (i) nonlinear electrical circuit (ii) linear electromechanical features. Validity of the model was done by conducting several experiments. A variational model for sensing and actuation behavior of IPMC is proposed by Buechler and Leo, (2007) and computationally analyzed using the Galerkin method. The frequency dependence of the piezoelectric coefficients is obtained from experiments. One-dimensional finite element model of a vibrating IPMC has been developed by Dogruer *et al.*, (2007) to analyze the effect of hydrodynamic loading on forced vibration. An experimental and numerical study was carried out by Kim *et al.*, (2007 a) on thrust generation of a vibrating IPMC strip in salted aqueous medium.

An analytical model has been developed for an IPMC actuator using Euler-Bernoulli's beam theory and the vibration response was studied by Zhang and Yang, (2007). Wallmersperger *et al.*, (2007) developed a computational model of transport and electromechanical transduction for IPMCs. A chemoelectrical coupling scheme is developed for transport model combining the Nernst-Planck equations for counter ions transport with the Poisson equation for the electric potential. A modified nonlinear version of the constitutive behavior proposed by Nemat-Nasser and Li, (2000) has been used to describe the actuation of IPMC actuator. A similar formulation has also been presented by Pugal *et al.*, (2008) and Porfiri, (2008). Pugal *et al.*, (2008) introduced a FE model to describe the time-dependent bending of a self-oscillating IPMC and Rayleigh damping was used to find out the viscoelastic behavior of IPMC. Porfiri, (2008) used matched asymptotic expansions to solve the nonlinear PDE and analyze the nonlinear electrical response of IPMCs.

Del Bufalo *et al.*, (2008) presented a three-dimensional framework to describe the mechanical actuation of IPMCs using theory of mixture. The model accounts for

osmotic pressure, hydraulic pressure and electrostatic forces, but only addressed the behaviour under quasi-static actuation. The vibration of a slender IPMC beam in a fluid medium is analyzed by combining an equivalent beam model and using classical Euler-Bernoulli cantilever beam theory by Brunetto *et al.*, (2008). In this model the interaction between IPMC and water is described by using the concept of hydrodynamic functions. Anton *et al.*, (2008) proposed an equivalent beam model for IPMC actuator to describe the non-uniform bending at large deflection in a cantilever configuration.

In order to understand the thrust generation mechanism of IPMC actuator, the hydrodynamics of a vibrating cantilever IPMC along its fundamental mode shape in an aqueous environment has been numerically analyzed by Abdelnour *et al.* (2009). Alici, (2009) proposed a model in a cantilever configuration to determine large and nonlinear bending deflection in longitudinal and transverse directions and to estimate the effective modulus of elasticity of conducting polymer actuators. Porfiri, (2009) developed a physics-based model relies on multiphase mixture theory for sensing and actuation of IPMCs undergoes small deflection. Truong *et al.*, (2010) proposed a nonlinear black-box model, based on the recurrent multi-layer perceptron neural network and self-adjustable learning mechanism to measure the tip displacement. McDaid *et al.*, (2010) proposed an electromechanical model which predict the actuation response of an IPMC actuator including tip displacement and blocked force accurately treating as large deflection bending problem. The model also takes into account the effect of nonlinearities at low frequencies. The model consists of three stages (i) nonlinear electric circuit to predict the current absorbed by the polymer (ii) electromechanical coupling stage for converting electric energy into mechanical energy and (iii) mechanical beam model to find out the stress generated along the length of the IPMC. Theoretical results are validated with experimentally obtained results. The proposed model overcomes the short comings of simple beam model with small deflection.

Lee *et al.*, (2011) developed a concentrated ion boundary layer model using finite element analysis to analyze the bending behaviour of platinum based IPMC actuator. The computational tip deflection and blocking force data are verified with

the experimental data. Gong *et al.*, (2011) developed a FE model coupled with a lumped RC circuit model to simulate the electro-mechanical response of IPMC actuator under an external input voltage.

1.3.4 Effect of Water Content on the Actuation Performance

Several studies have been carried out to investigate the effect of water content on the actuation performance of IPMC. Nemat-Nasser, (2002); Nemat-Nasser and Wu, (2003) developed a physics-based model to investigate the change in stiffness with hydration level and compared with experimental data. The results show that, the stiffness of IPMC is significantly influenced by the hydration level. Electrochemical and thermal experiments were conducted to investigate the effect of water content on the actuation performance of IPMC by Lee *et al.*, (2005a). Using a cyclic voltamogram they showed that, the electric current is proportional to the water content in the IPMC under the same electric potential. A model has been proposed by Enikov and Seo (2005) and experimental investigation is carried out at high voltage ($>1.2\text{V}$) to find out the effect of current on the initial water content and displacement. Experimental investigation has been carried out by Yeh and Shih (2010) to find out the effect of water content on the actuation performance of IPMC for 7200 s actuation time with AC voltage. They showed that, with decrease in initial water content the actuation performance of IPMC decreases, and maximum deformation of the IPMC is obtained when the initial water content is around 70%.

1.3.5 Applications

Several characteristics of the IPMC actuator like; low operating voltage, large bending deformation, ease of miniaturization, and light weight, distinguishes them from the other class of active actuators. The low actuation potentials can be easily implemented by simple control circuits and well suited to battery-operated devices. As a polymer, IPMC can be easily designed into various shapes; their properties can be tailored to a specific application by following different manufacturing process. The most attractive feature of IPMC is their ability to mimic the biological muscles with high fracture toughness, large deformation, and inherent vibration damping.

The use of IPMC actuators could eliminate the need for gears, bearings, and other components that complicate the construction of micro-robots and other similar devices, which reduce the cost and the weight. An overall review of how the materials are commercializing are discussed by Ashley, (2003). Specific industrial and medical applications also discussed in Shahinpoor and Kim, (2005).

Robotic Applications: Shahinpoor *et al.*, (1998) applied IPMC in the development of a finger-like micro-gripper. Lumia and Shahinpoor, (1999) designed and developed a multi-finger gripper at UNM Artificial Muscles Research Institute using IPMC. Lee *et al.*, (2006b) fabricate a thick IPMC films by hot-pressing several thin IPMC films and developed IPMC actuators which were used for developing artificial fingers. Jain *et al.*, (2009) developed a three fingered Remote Center Compliance (RCC) micro gripper using IPMC to facilitate insertion of peg-in hole for micro assembly operation. Hanson *et al.*, (2001) emulated facial anatomy with a robotic humanoid head using dielectric elastomer called electrostatically stricted polymers (ESSP) at University of Pisa, Italy. Arena *et al.*, (2006) uses IPMC actuator to fabricate wormlike robot, which is controlled by cellular neural networks (CNNs).

Guo *et al.*, (2000) developed biomimetic fish-like propulsion using Ionic Conducting Polymer Film (ICPF) actuator for an underwater microrobot swimming structure in water. Laurent and Piat, (2001) uses IPMC actuator to develop a swimming microrobot propelled by the beating of two fins. Jung *et al.*, (2003) also applied ionic polymer metal composite (IPMC) actuator to develop a wireless tadpole. Kim *et al.*, (2005) demonstrated a biomimetic, undulatory tadpole robot that used IPMC as actuator. Guo *et al.*, (2006) developed a fish-like S-shape swimming micro robot, using IPMC actuator for body and tail fin propulsion. A biomimetic underwater walking microrobot utilizing 6 ICPF actuators as legs was developed by Zhang *et al.*, (2006), 8 IPMC actuators as legs by Guo *et al.*, (2008), and 10 IPMC actuators as legs of fingers by Shi *et al.*, (2010). Tangorra *et al.*, (2007) used conducting polymer actuators based on polypyrrole to develop biorobotic artificial pectoral fins of the bluegill sunfish. Trivedi *et al.*, (2008) discusses robot design, modelling challenges, fabrication, and control techniques of biologically inspired

soft robots. Yeom and Oh, (2009) fabricate a biomimetic jellyfish robot using IPMC actuators. Zhang *et al.*, (2010) developed biologically inspired artificial fish using flexible matrix composite (FMC) actuators. Chen *et al.*, (2011) built a free-swimming robotic batoid ray using two IPMC actuators as pectoral fins. Shahinpoor *et al.*, (1998); Yim *et al.*, (2007); Lu *et al.*, (2008) demonstrate potential of IPMC actuator as artificial muscles. Akle and Leo, (2003) studied the possibilities of use of IPMC in flapping Micro Air Vehicle (MAV).

Medicine and Biotechnology: Biomedical applications favour ionic polymers as the base material Nafion/Flemion is biocompatible [Gerritsen *et al.*, (2000)]. This biocompatibility makes the ionic polymer transducer suitable for application inside the human body for many applications. Some of the major applications are discussed in this section. Guo *et al.*, (1994, 1995); Sewa *et al.*, (1998); Onishi *et al.*, (2000); Fang *et al.*, (2007) proposed and developed micro active catheter system using IPMC actuator. Keshavarzi *et al.* (1999) proposed IPMC as sensor to measure blood pressure, pulse rate and rhythm. Mudarri *et al.*, (2004) reported that IPMC sensor can be used to monitor bone healing progress. Feng and Tsai, (2009) demonstrated that IPMC actuator can be used in microendoscopic ocular surgery. Li *et al.*, (2011) developed helical IPMC actuator, for biomedical active stents to treat cardiac disease.

Micro-Fluidic Applications: IPMC actuators/sensors are suitable for micropump applications since it can be operated in aqueous environment with low applied potential and can produce large stroke volumes along with controllable flow rates [Lee *et al.*, (2005)]. Hattori *et al.*, (1992) proposed and developed micropumps using polymer gel actuators. Guo *et al.*, (1997) uses ICPF actuators to develop a prototype model of a micro pump. Guo *et al.*, (1999) further developed a capsule micropump (13 mm diameter, 23 mm length) using ICPF actuator as servo actuator. Lee *et al.*, (2005) investigated the application of IPMC actuator in design of micropump. Lee and Kim, (2006) outlined the design method of an IPMC actuator-driven valve-less micropump. Nguyen *et al.*, (2008) proposed a design, fabrication and experimental

characterization of a flap valve IPMC micropump with a flexible supported diaphragm.

Planetary Application: Ionic polymer metal composite (IPMC) actuators are also being used in the space systems such as, dust accumulation over the optical instruments of solar cells. Bar-Cohen *et al.*, (2000) proposed the idea of using IPMC actuator as the planetary dust wiper, which can remove dust from surfaces.

Other Applications: Tailoka *et al.*, (2003) developed and assembled a device to measure the humidity in a chlorine-air atmosphere using IPMC sensor. Bandopadhyaya *et al.*, (2007, 2008) proposed IPMC actuator as an active vibration damper for vibration control of a large deflected flexible link. Bandopadhyaya *et al.*, (2009) uses IPMC actuator as a partially compliant rocker in a four-bar mechanism for work volume generation.

1.4 Motivation of the Present Work

Natural muscles are highly evolved material systems that enable the production of large deformations by repetitive molecular motions. There is currently no widely used technology to replace or stimulate muscle, which is a strong motivation for the proposed research work.

So far, IPMC has proved research-worthy and with tremendous potential applications in biomimetic and medical application. A key advantage is that it requires very low actuating voltage of around 3-5V, while PZT actuator normally requires around 200-500V. This eliminates the extra mass and cost associated with the power electronics, making IPMC a light weight alternative. Again, PZT actuators are manufactured with relatively small dimension due to high brittleness and may lead to breakdown under fatigue loading. Another advantage of IPMC over PVDF is that the latter is more compliant and sensitive to the change of applied strain. Magnetostrictive materials undergo huge hysteresis loss and difficult to fabricate in actuator form and as well as in embedding. Unlike other smart materials, ionic polymer metal composites can undergo large displacements at relatively low

voltages (1-3V). They also exhibit relatively fast response and can be operated very well in aqueous environments.

However, there are very less commercial applications of ionic polymer metal composites as actuator or sensor till date. One reason could be that, the technology is still new and the working mechanism and further improvement on it is not well explored; another reason is that the manufacturing of IPMC is very expensive. Reason behind the high manufacturing cost of these composites is due to the high cost of the precious metals such as platinum (Pt) or gold (Au) that are typically used as the surface electrode. Considering the range of applications of these polymers, have great future potential to be used in many industrial, medical, space and other research applications, but are limited due to the high manufacturing cost. This motivates to use non-precious metal silver (Ag) as surface electrode which has shown to be research worthy for improving the performance of IPMC. This reduces the cost of fabrication of IPMC to be used as an actuator and sensor for biomedical and other artificial prosthetic applications.

Several studies have been carried out to find out the effect of water content on the actuation performance of IPMC actuator. Researchers established the fact that an IPMC actuator dehydrates continuously in open environment reduces the initial water content with applied electric potential, and affect significantly the actuation performance. Till date, limited studies have been carried out and hence, a detailed research is required to explore the effect of dehydration on actuation performance. This motivates to develop a simple but robust empirical model using Cobb-Douglas production method from the experimental data to predict the loss of water content in terms of input voltage and time in the open working environment.

The precise predication of stable and unstable zone for an applied excitation voltage and frequency is of importance in many micromechanical systems utilizing IPMC as the actuator. Mostly all of the proposed mathematical models on IPMC actuator are based on linear modeling. A linear dynamic model thus will no longer be valid for IPMC actuator having low thickness and thus more flexible. Literature shows a little research has been carried out to find out the nonlinear vibration characteristics of IPMC actuator, excited by a small alternating electric potential.

Though development of a comprehensive model is still going on, factors such as dynamic response and nonlinearities have received less attention in the previous work. This motivates the development of a nonlinear dynamic model that is capable of better describing the behaviour of ionic polymer metal composite actuator during actuation.

Active materials are nowadays utilized successfully for interventional diagnosis and therapy for their biocompatibility, suitability for smooth and safety operation. Active catheter is one such mechanism where active materials such as shape memory alloy (SMA), electroactive polymers (EAPs) and piezoelectric materials are successfully used over the few years. This motivates to develop an active novel variable length catheter mechanism system using the fabricated IPMC actuator for *in-vivo* oral and body inspection or scanning.

1.5 Objectives of the Present Work

The specific objectives of this research work are outlined below along with the specific contribution to the field of ionic polymer metal composite actuator. The main objectives are stated below.

- ❖ Fabrication and characterization of ionic polymer metal composites (IPMCs) using non-precious metal silver (Ag) as surface electrode.
- ❖ Investigation and calculation of loss of water content (loss-factor) in the open working environment under potential.
- ❖ Analyze the transient and steady-state response of IPMC actuator under potential gradient and to demonstrate the effect of dehydration on vibration response both theoretically and experimentally for DC input voltage.
- ❖ To develop a nonlinear governing equation of motion and a closed form expression for the response amplitude and detuning parameter for obtaining the nonlinear frequency response curve for any IPMC actuator. The developed simple expression can be used for finding the response of the system instead of solving the temporal equation of motion which is time consuming and requires more memory space.

- ❖ To develop an active novel variable length catheter mechanism using the fabricated IPMC actuator for *in-vivo* oral and body inspection or scanning.

1.6 Organization of the Thesis

The thesis consists of seven chapters, and is organized as follows:

First chapter deals with a brief introduction and actuation mechanism of ionic polymer metal composites followed by literature review. Motivations of the present work, objectives and thesis organization are outlined in this chapter. The details of fabrication procedure, thermal, electrical and mechanical characterization of a novel silver coated IPMC actuator is discussed in chapter two. The experimental results are studied and compared with the literatures in this chapter.

In chapter three, a simple but robust mathematical model is developed from the experimental data to predict the loss of water content in terms of input voltage and time in the open working environment. Variable parameters pseudo-rigid body modelling (VPPRBM) approach is used to estimate and investigate the effect of dehydration on tip position and bending response of IPMC actuator.

In chapter four, theoretical modelling of the actuator is performed using multi-mode approximation of a fixed-free Euler-Bernoulli beam and the governing equation of motion of the actuator has been derived by using the extended Hamilton's principle incorporating loss due to dehydration. Several experiments are conducted to investigate the vibration characteristics and the results are validated and found to be in good agreement with the theoretical results.

In continuation, nonlinear vibration characteristics of a cantilevered IPMC actuator are analyzed under alternating electric potential in chapter five. The theoretical results have also been experimentally validated. Chapter six, deals with an application of IPMC actuator in active catheter mechanism. An active variable length catheter system has been developed and analyzed. Finally, chapter seven provides the summary of the contributions of the thesis and scope for future work.

Fabrication and Characterization of Silver-electroded Ionic Polymer Metal Composite Actuator

2.1 Introduction

The current state-of-the-art procedure to fabricate the artificial muscle i.e., ionic polymer metal composite (IPMC) with superior performance is expensive as the surface metal electrodes are made of with precious metals like platinum or gold. There is also serious limitation of arresting moisture content which results in poor performance during experiment. It is expected that changing processing and operating conditions may reduce the cost of fabrication and also improve the performance of IPMCs. In the recent year, low cost IPMC based on non-precious metal like silver has also been developed. However, limited success has been achieved for successful fabrication of Ag-IPMC to obtain superior and reliable performance compared to other noble metal based IPMCs. The proposed research objective is to use non-precious and cost-effective metal silver (Ag) for surface electrode which has proved to be research worthy for improving the performance of IPMC. This also reduces the cost of fabrication of IPMC to be used as an actuator and sensor for biomedical and even in artificial prosthetics application. In this chapter the followings have been discussed:

- Fabrication of ionic polymer metal composite (IPMC) with silver (Ag) as the surface electrode.
- Characterization of thermal, electrical and mechanical properties of the developed Ag-IPMC.
- Comparative study on performance of the developed Ag-IPMC with other noble metal based IPMCs.

Chemical decomposition method is adopted to fabricate the proposed pure Ag-electroded IPMC. Several steps of trial-and-error method are carried out for optimizing the process parameters and fabrication procedures as detailed procedure was not mentioned in the previously published work [e.g., Chen *et al.*, (2009) and Tamagawa *et al.*, (2003)]. Further, the composition of chemicals used in the literatures Chen *et al.*, (2009) and Tamagawa *et al.*, (2003) are different from those used in this work.

Microscopic and morphological analysis, X-ray diffraction (XRD) analysis, thermo-gravimetric analysis (TGA), differential scanning calorimetry (DSC) tests, dynamic-mechanical analysis (DMA) tests and thermo-mechanical analysis (TMA) are carried out to characterise the fabricated IPMC and the observations are discussed in the following sections. Under DC potential the bending actuation test is conducted in a fixed-free configuration for the developed IPMC actuator. Experiments are carried out to measure the tip force, surface resistance, solvent loss, hydration level and operational life of the actuator. Micro-tensile test of the specimen is carried out to ascertain the stress-strain relationship and comparison is made with the base polymer Nafion. Also experiments are conducted to study the vibration characteristics of the actuator using Laser Vibrometer.

2.2 Fabrication of Ag-IPMC

As discussed in the chapter one, there are several fabrication techniques like electroding by chemical plating/decomposition, sputter coating, physical vapour deposition (PVD) and solution casting method to fabricate IPMC. Electroding by chemical plating/decomposition technique is mostly acceptable method for fabricating IPMCs and this method is used here to fabricate the proposed pure Ag-IPMC. The materials required for fabricating Ag-IPMC are: Nafion-117 as base polymer, (purchased from Ion Power, Inc, New Castle, DE 19720 USA); silver nitrate (AgNO_3) as silver salt and dilute ammonia solution (NH_3) to prepare diamminesilver (I) hydroxide, $\text{Ag}(\text{NH}_3)_2\text{OH}$ -silver complex solution used for providing the cations (Ag^+); sodium hydroxide (NaOH) for providing Na^+ cations into the membrane; dextrose anhydrous GR ($\text{C}_6\text{H}_{12}\text{O}_6$), as reducing agent to reduce

the silver cations to silver metal; hydrochloric acid (HCl), 2N solution as a cleaning agent; and lastly deionized/distilled water for cleaning, preparation of chemical solutions and storing the IPMC. The flowchart for the fabrication process is given in Figure 2.1 and the step-by-step procedure is described as follows.

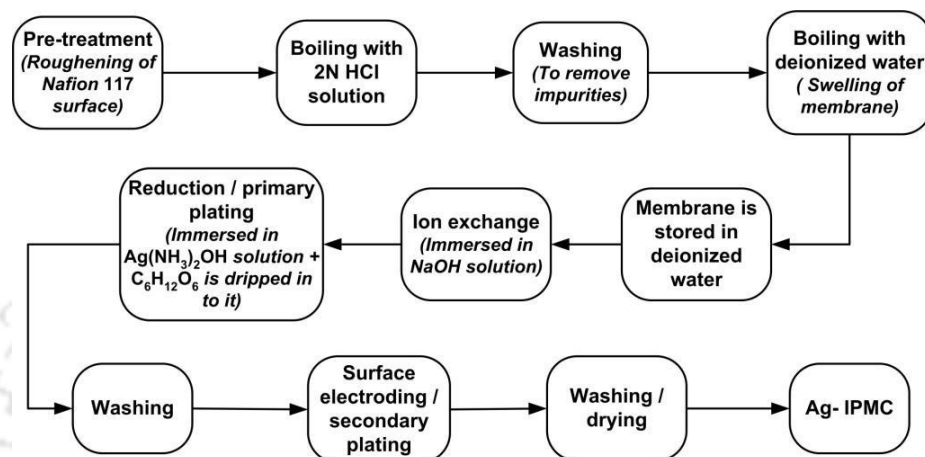


Figure 2.1 Flow chart to fabricate pure Ag-IPMC membrane by chemical decomposition technique.

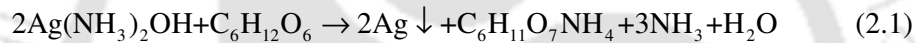
Step 1: Surface roughening and chemical cleaning of the membrane

- Surface preparation: Roughening of both surfaces of Nafion membrane is done by using metallography silicon carbide grinding paper of grain fineness P800. The process increases surface area i.e. open up more sites for arresting the silver ions.
- Treatment with HCl: Nafion membrane is boiled in dilute hydrochloric acid (2N solution) for 30 min to remove impurities from the membrane. Subsequently, rinsing of the membrane is done with deionized water.
- Treatment with water: Membrane is boiled in deionized water for 30 min to remove acid and for swelling. The membrane is then stored in deionized water till the initiation of next step.

Step 2: Adsorption: A solution of sodium hydroxide (NaOH) is prepared with concentration of (0.50 mol/l). The quantity of Na^+ cations absorbed depends on the charge of the complex solution and the absorptivity of the cation into hydrated Nafion. The solution is stirred for 1h with the membrane in it and subsequently the membrane is kept within it for 6h to 8h to ensure adequate

diffusion of Na^+ into the membrane. A complex salt solution of diamminesilver (I) hydroxide is prepared by adding ammonia solution drop by drop into the silver nitrate solution with a concentration of (0.15 mol/l). The solution of diamminesilver (I) hydroxide, $\text{Ag}(\text{NH}_3)_2\text{OH}$ with the membrane is stirred for 1h and subsequently the membrane is kept immersed within the solution for 6 h to let $\text{Ag}(\text{NH}_3)_2^+$ diffuse into the membrane.

Step 3: Primary plating (reduction): An aqueous solution of dextrose anhydrous GR (reducing agent) with concentration (0.0888 mol/l) is added drop by drop into the solution of diamminesilver (I) hydroxide and is stirred for few min. The process results in deposition of shiny layer of fine silver particles on both surfaces of the membrane. The reaction that takes place is given by [Chen *et al.*, (2009)]:



The membrane is then put into the sodium hydroxide solution with a concentration of (0.50 mol/l) and is stirred in the ultrasonic-clean-bath for 30 min at 20-25°C to clean any loosely deposited Ag particles on the surface. The process initiates a black silver deposition near the surface of the material. The sample is then stored in deionized water till the next stage is performed.

Step 4: Secondary plating (developing): Additional amount of silver is deposited on the initial layer by developing process. The membrane is immersed into the dextrose anhydrous GR solution for approximately 2h at room temperature. The membrane is then further soaked with diamminesilver (I) hydroxide solution for approximately 3h at room temperature. $\text{C}_6\text{H}_{12}\text{O}_6$ solution is then dripped into the beaker containing the membrane for several times with continuous stirring. The process ensures another silver layer is formed on the surface of the Nafion membrane. The membrane is then rinsed with deionized water and stirred in the ultrasonic cleaning bath to remove any loosely deposited Ag particles on the IPMC surface. Final IPMC strip is obtained by trimming the membrane from all sides to avoid any shorting between two surfaces.

All the above steps are also illustrated pictorially in Table 2.1. The time required for the whole fabrication process of the proposed work is around 24-25h, which is almost 50% less than the time required for the existing fabrication of Pt-IPMC [Shahinpoor and Kim (2001)]. The cost of fabrication for a $40 \times 40 \times 0.2 \text{ (mm}^3\text{)}$ Ag-IPMC is around Rs. 2500.00 (50 USD), which is much less compared to the platinum / gold based IPMC (The cost of a set of 4 polymeric actuators/sensors/transducers/energy harvesters elements of size $10 \times 40 \times 0.2 \text{ (mm}^3\text{)}$ is 989 USD i.e., one IPMC actuator is around 250 USD) [http://www.environmental-robots.com/products.html; date: 30/10/2011]. Hence, for an Ag-IPMC of size $10 \times 40 \times 0.2 \text{ (mm}^3\text{)}$ the fabrication cost including the material cost is twenty times less than that of the Pt/Au based IPMCs.

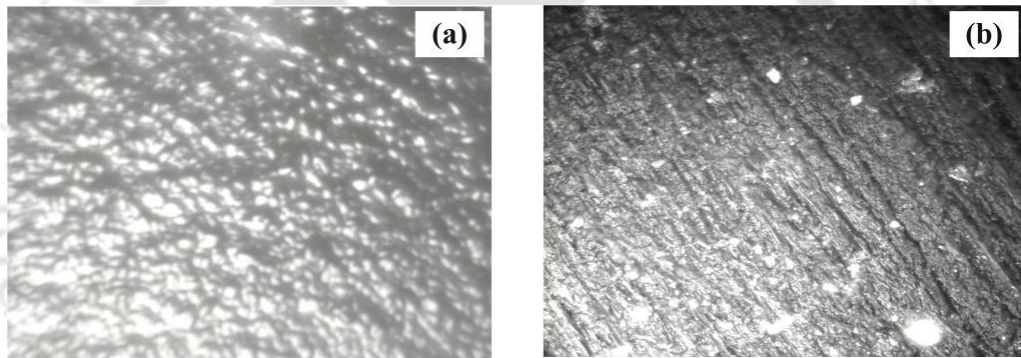


Figure 2.2 Surface micrograph of the Nafion-117 membrane (a) pure state (b) after roughening.

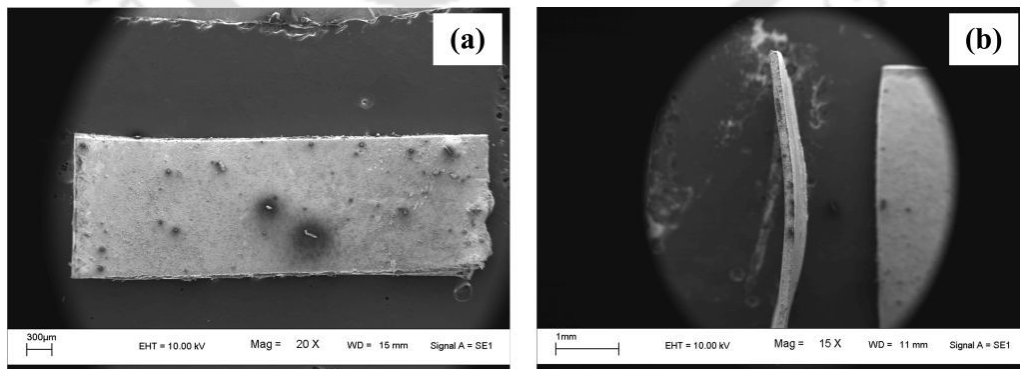
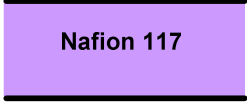
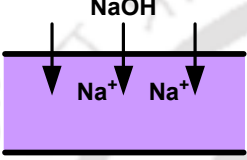
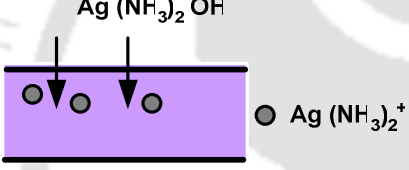
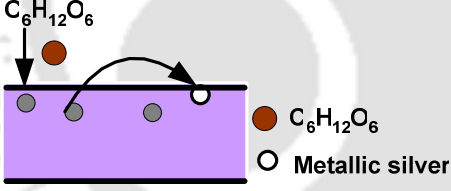
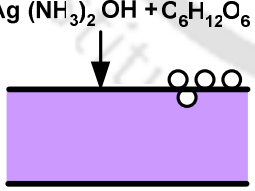
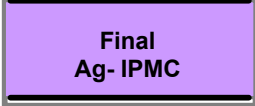


Figure 2.3 SEM images of fabricated IPMC actuator (a) surface (top - view) (b) along the thickness (side-view).

Table 2.1 Fabrication processes of pure Ag-IPMC actuator.

Step	Processes	Details
i		Pre-treatment Roughening of Nafion-117 membrane surface to increase the surface area for arresting the silver ions.
ii		Adsorption Exchanging of Na^+ and $\text{Ag}(\text{NH}_3)_2^+$ ions into the Nafion membrane.
iii		
iv		Primary plating The process ensures silver cations penetrate into the Nafion membrane in depth.
v		Secondary plating Additional silver layers are developed on the primary plating surface to improve the electrical conductivity of the electrodes.
vi		Final Ag-IPMC.

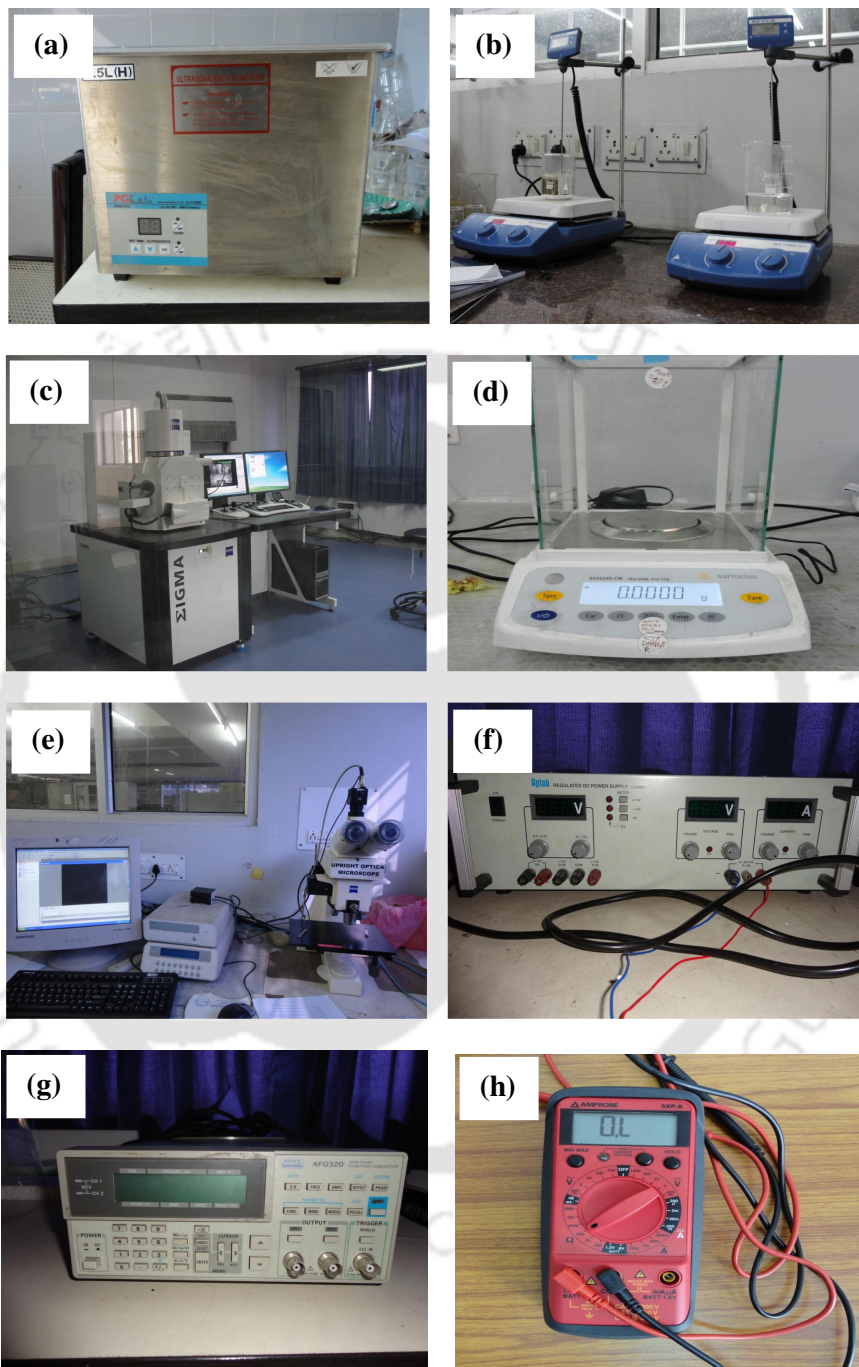


Figure 2.4 Various equipments used in the experiments (a) Ultrasonic cleaning bath (b) Magnetic stirrer (c) SEM (d) Digital balance (e) Microscope (f) DC power supply (g) AC power supply (h) Multimeter.

Figure 2.2(a) and (b) show the surface micrograph of the Nafion-117 membranes at pure state and after the roughening, respectively with an amplification ratio of 40 times (Upright optical microscope, Make: Carl Zeiss, Model: Axiotech-100 HD). The effect of roughening is clearly observed in Figure 2.2(b). Figures 2.3(a) and (b) show the top and side surface images taken by scanning electron microscope (SEM) of the fabricated IPMC respectively, with amplification ratio of 20X and 15X, respectively. The images are taken by using LEO 1430VP Scanning Electron Microscope supplied by LEO Electron Microscopy Inc. Figure 2.4 shows some of the equipments used for experimental purpose to characterize the properties of the IPMC and Nafion.

2.3 Characterization of Ag-IPMC

2.3.1 Morphological and Microstructure Analysis

The surface morphology of the IPMC is studied using SIGMA Field Emission Scanning Electron Microscope (FE-SEM) (Make: Carl Zeiss NTS GmbH). A sample specimen of 5 mm x 5 mm is prepared and is examined under high voltage (10 kV) with 1.17 kX times magnification. Figure 2.5(a) illustrates the SEM based morphological structure of the fabricated Ag-IPMC surface. Energy dispersive X-ray (EDX) analyzer is equipped with scanning electron microscope (Make: LEO 1430VP), analyze the chemical composition of the sample. The micrograph shows densely packed and adequate dispersion of silver particles on the surface and their interfacial bonding with Nafion membrane. Figure 2.5(b) demonstrates the EDX analysis of the surface. The result clearly shows the presence of Ag particles over the surface of the IPMC. Figure 2.6(a) shows the cross-sectional morphology of the IPMC membrane. It is observed that the Ag particles penetrate the surface of the Nafion membrane up to 8-9 μm . Figure 2.6(b) shows the energy-dispersive X-Ray (EDX) analysis of a section of the IPMC actuator. The results demonstrate the mixture of Ag particles and Nafion membrane compounds over the cross-section of the IPMC. Morphological study reveals that the average diameter of the Ag particle is around 0.5-0.6 μm . While cracks have been observed in the IPMC developed by

Chen *et al.*, (2009) with a magnification of 400 times, however no cracks have been observed in the developed IPMC even under 1.17kX magnification. Hence the proposed method is proved to be efficient for fabrication of Ag-IPMC.

To further investigate the penetration depth of the silver and its derivatives composition on the Ag-electrodes in the cross-section along the thickness, EDX is carried out at three different points to a few micron depth starting from outside to the inside of the surface as shown in Figure 2.6(a) and the results are summarized in Table 2.2. This shows the high concentration of Ag on the outer surface gradually decreases towards inside of the Nafion surface.

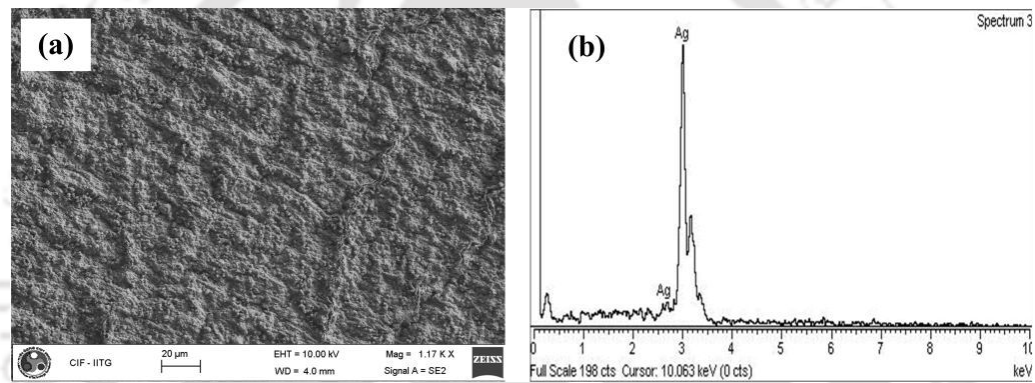


Figure 2.5 (a) SEM micrograph (b) EDX of the surface of the fabricated IPMC.

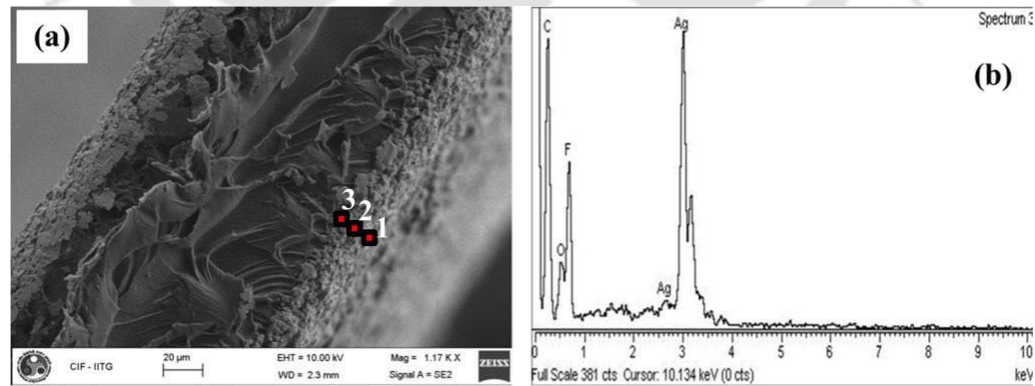


Figure 2.6 (a) SEM micrograph (b) EDX of the cross-section of the fabricated IPMC.

Table 2.2 EDX analysis of an Ag-IPMC sample as shown in Figure 2.6 (a).

Points	C	O	F	Ag	Total
1	5.62	2.07	27.53	64.78	100
2	14.78	7.80	18.91	58.51	100
3	18.67	-	48.29	33.04	100
Maximum	18.67	7.80	48.29	64.78	
Minimum	5.62	2.07	18.91	33.04	

2.3.1.1 X-ray Diffraction Analysis (XRD)

The crystal structure of the Nafion and IPMC membranes are investigated using a Bruker D8 Advance X-ray diffractometer (XRD, Cu K α radiation at 40 kV and 40 mA) instrument. Figure 2.7 shows both the XRD patterns of Nafion and IPMC membrane. Three major characteristics diffraction peaks of Ag-IPMC are observed at $2\theta = 38.15^\circ$, 44.3° , and 64.45° which are corresponding to the crystal faces of (111), (200), and (220).

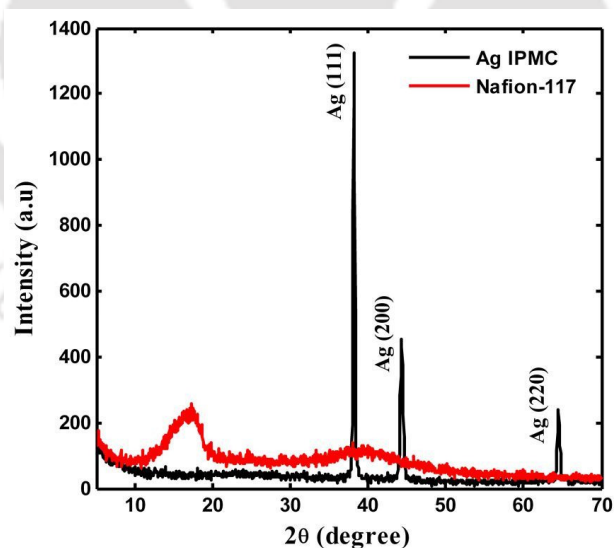


Figure 2.7 XRD patterns of the Nafion-117 and Ag-IPMC membrane.

The observation confirms that deposited silver film is polycrystalline in nature without any unwanted phase formation. In fact, it shows the characteristics of face-

centred cubic (FCC) structure which validates that silver-coated IPMC have perfect conductivity property. The characteristics diffraction peak of Nafion membrane at $2\theta \approx 18^\circ$ is in good agreement with the previous results of Lee *et al.*, (2005a).

2.3.2 Thermal Analysis

Investigation of thermal stability of electro-active polymers is indispensable for various applications as sensor and actuator in an environment subjected to temperature fluctuation. Thermo-gravimetric analysis (TGA) and differential scanning calorimetry (DSC) analysis are carried out to investigate the thermal behaviors and also the effect of temperature on the Nafion and IPMC membranes. TG and DSC measurements are carried out on a NETZSCH STA 449 F3, Jupiter® instrument. Nafion and IPMC films are cut into small pieces to provide a suitable sample size of about 5-10 mg. TG and DSC curves are recorded on Nafion in acidic form (Nafion H^+) and IPMC membrane at a heating rate of $10^\circ\text{C}/\text{min}$ under Argon atmosphere (60 ml/min) from room temperature to 650°C (TG curves) and 550°C (DSC curves), respectively. Figure 2.8 shows the photograph of the DSC/TGA instrument.



Figure 2.8 Photograph of the DSC / TGA instrument.

2.3.2.1 Thermo-Gravimetric Analysis (TGA)

Both pure Nafion and IPMC membranes in dry state are used for thermo-gravimetric analysis (TGA). Figure 2.9(a) and (b) show the TG curves of Nafion-117 and IPMC membranes respectively. The solid line represents the percentage of mass loss, while the dashed line is the first derivative (DTG) curve. The TG curve shows different profiles for Nafion and IPMC membrane. Nafion-117 is a perfluorosulfonate ionomer consists of a polytetrafluoroethylene (PTFE) backbone with side-chains terminated with a sulfonate group. In tune of the work of Almeida and Kawano, (1999) the decomposition of Nafion and IPMC membranes can be categorized into 4 different stages with temperature which are listed in Table 2.3. The results obtained for Nafion membrane (Figure 2.9(a)) is in good agreement with the experimental results of Stefanithis and Mauritz, (1990).

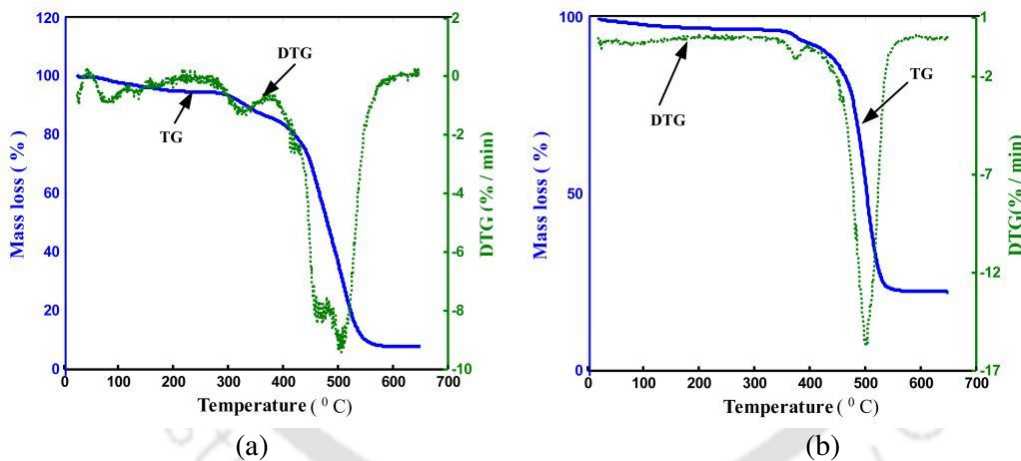


Figure 2.9 TG (solid line) and DTG (dotted line) curves of (a) Nafion-117 (b) IPMC membrane.

Figures 2.9 (a) and (b) clearly demonstrate that due to presence of silver particles the thermal stability of IPMC improved significantly than pure Nafion. It is also observed that at higher temperatures, (470-575°C) the Nafion membrane as well as the silver particles starts degradating. Table 2.3 summarizes the percentage of mass loss and its causes at various temperatures. It is also observed that Nafion and IPMC membranes are thermally stable below 270°C and 350°C, respectively.

Table 2.3. Percentage of mass loss for Nafion-117 and Ag-IPMC.

Temperature range (°C)	Mass loss, Nafion-117 (%)	Mass loss, Ag-IPMC (%)	Observation
25-290	5.9	2.89	Loss of moisture
290-400	10.16	3.83	Desulfonation
400-470	26.25	11.09	Side-chain decomposition
470-575	49.63	59.2	Backbone decomposition
575-650	0.247	0.19	Residues

2.3.2.2 Differential Scanning Calorimetry (DSC) Analysis

Figures 2.10(a) and (b) demonstrate DSC thermographs of Nafion and IPMC membranes, respectively. DSC thermograph for Nafion (Figure 2.10(a)) shows a strong endothermic peak near 88°C as glass transition temperature (T_g), and another peak near 143°C as crystallation temperature (T_c). Thermal transition above 200°C may be associated with the thermal degradation of Nafion membrane. The high exothermic peak near 332°C indicates the melting point temperature (T_m) of the Nafion membrane.

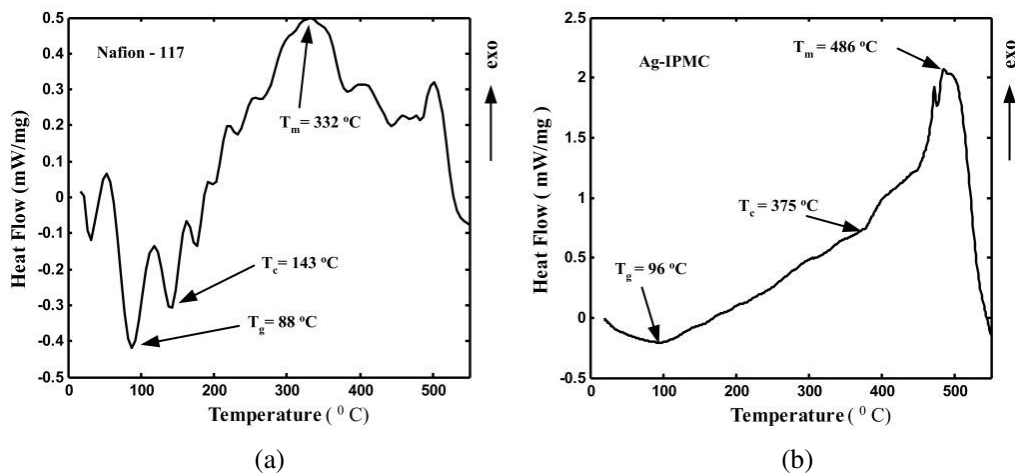


Figure 2.10 DSC thermograph of (a) pure Nafion membrane (b) Ag-IPMC.

As shown in Figure 2.10(b) the (T_g), (T_c) and (T_m) temperatures for IPMC are observed as 96°C, 375°C, and 486°C, respectively. It is anticipated that, the increase in T_g , T_c , and T_m of IPMC could be due to the presence of silver particles within the Nafion membrane.

2.3.2.3 Dynamic Mechanical Analysis (DMA)

Dynamic-mechanical analysis (DMA) is carried out for the Nafion and IPMC membranes in dry condition for studying viscoelastic properties of both the polymer and composite. The test is conducted in open environment in tensile mode in an EXSTAR TMA /SS 6000, SII Seiko Instrument Inc. instrument as shown in Figure 2.11. The frequency of applied load is taken as 0.1 Hz while the heating rate is kept constant at 5°C/min.

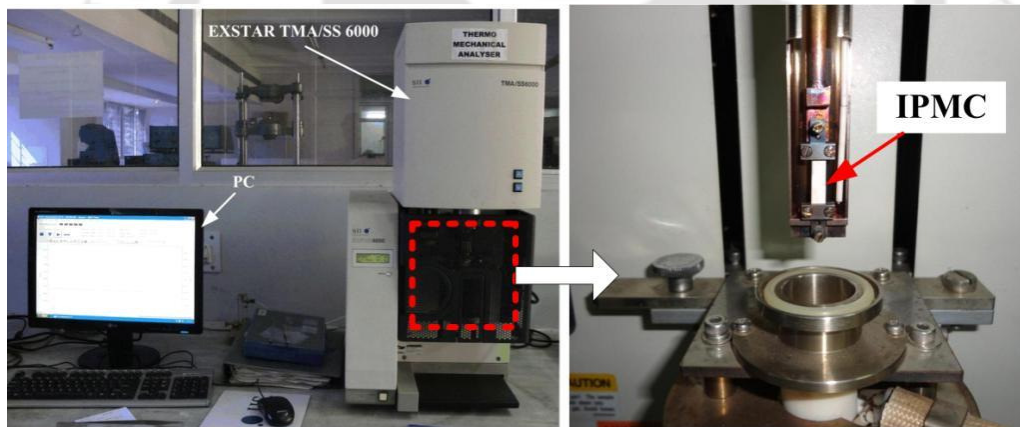


Figure 2.11 Photograph of the DMA/TMA instrument.

Figure 2.12 (a) demonstrates the change in storage modulus (E') of both Nafion and IPMC with temperature. It is observed that initially the storage modulus of IPMC increases up to $3.2 \times 10^8 \text{ N/m}^2$ with increase in temperature up to 25°C and then decreases to $2.5 \times 10^8 \text{ N/m}^2$ at temperature 45°C. With further increase in temperature, it further increases to a maximum of $3.5 \times 10^8 \text{ N/m}^2$ after which it decreases. The glass transition temperature (T_g) for Nafion and IPMC membrane is obtained from the peak value of $\tan \delta$ and it is found to be 142.8°C and 210.8°C, respectively. Park *et al.*, (2007) carried out the DMA test for Nafion, Pt, Au, Pd and IL-Pt IPMCs at 1Hz in tensile mode and the T_g values are found to be 112.5°C, 121.4 °C, 157.7°C,

118.9°C and 70.1°C, respectively. Kim *et al.*, (2010) also conducted DMA test for Nafion, Pt, Au IPMCs in a tensile mode at 1 Hz with heating rate 2°C /min and T_g values are found as 112.15°C, 121.35°C and 157.76°C, respectively. The DMA glass transition temperature (T_g) shows the substantial change in the rigidity of the polymer during a short span of temperature. It is observed that T_g for Ag-IPMC shifted to a higher temperature due to the presence of silver particles on the Nafion membrane.

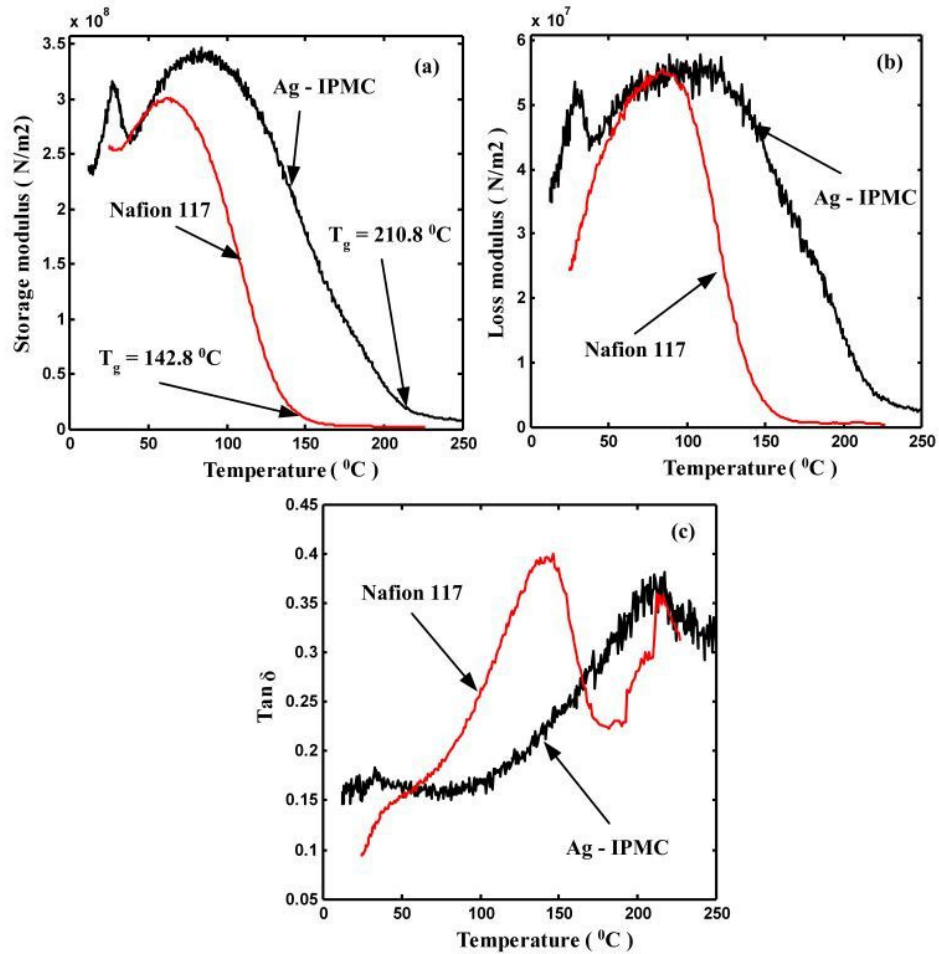


Figure 2.12 (a) Storage modulus (E') (b) Loss modulus (E'') (c) Tan δ as a function of temperature for Nafion and Ag-IPMC at 0.1 Hz (heating rate 5°C/min).

Figure 2.12(b) demonstrates the variation of dynamic loss modulus (E'') for both Nafion and IPMC with temperature. It is anticipated that initial rise in the loss modulus is due to the chain lengthening effect which causes the viscosity to rise

with temperature. Figure 2.12(c) shows the $\tan \delta$ curve and is obtained by taking the ratio E''/E' , represents the damping characteristics of the samples. $\tan \delta$ indicates how efficiently a material losses energy due to molecular rearrangements. It is observed that Ag-IPMC shows a high storage modulus and a high loss modulus simultaneously. This results suggest that Ag-IPMC having improved stiffness and better thermal stability than pure Nafion.

2.3.2.4 Thermo Mechanical Analysis (TMA)

Thermal expansion data are collected on film's surface by thermo-mechanical analyzer (EXSTAR TMA /SS 6000, SII Seiko Instrument Inc.) using a macro-compression alumina probe as shown in Figure 2.11. Data are collected at a heating rate of $10^\circ\text{C}/\text{min}$ from room temperature to 300°C while the compression load is adjusted to 0.1 N. Calibration of the instrument is done using zinc standard and a sample of $7\text{mm} \times 6\text{mm}$ is prepared and tested on TMA and coefficient of thermal expansion (α) is obtained. The variation of α with temperature is shown in Figure 2.13.

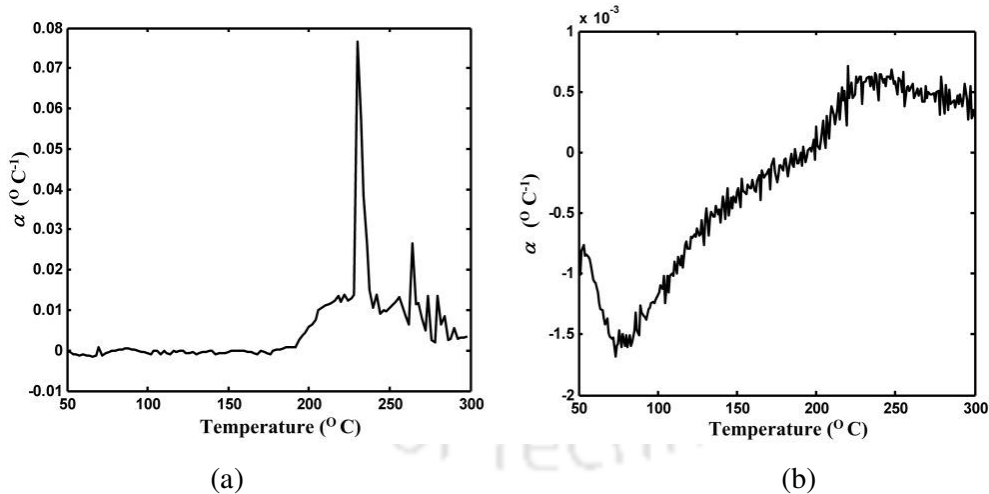


Figure 2.13 Variation of coefficient of thermal expansion (α) with temperature for (a) Nafion-117 film (b) pure Ag-IPMC membrane.

Figure 2.13(a) and (b) show the variation of coefficient of thermal expansion (CTE) of the Nafion film and Ag-IPMC membrane with temperature, respectively. It is understood from the figures that the addition of silver particles influences the CTE of the Nafion polymer film. The CTE of Nafion film (Figure 2.13(a)) is found to be

around $0.002^{\circ}\text{C}^{-1}$ from room temperature to 100°C , which is in good agreement with the results of Tang *et al.*, (2008). As shown in Figure 2.13(b), it is observed that the CTE of Ag-IPMC shows a negative value from 50°C to around 80°C and then a positive value from 200°C to around 300°C . The range between 80°C to 250°C is very important as the CTE gradually increases from negative to positive value. For a Pt based IPMC this range is found to be 100°C to 150°C [Tang *et al.*, (2008)]. This shows that the proposed Ag-IPMC has opposite thermal responses at different temperatures i.e. the IPMC exhibits bi-material property responding to temperature such that it shrinks when the temperature is lower than 80°C and scales up by sudden expansion when the temperature is beyond 80°C . In addition, when the temperature increases the moisture content in the surface of IPMC film is reduced due to dehydration, weakening the outer layer of silver coating. It is anticipated that, with the applied voltage the bending deformation of IPMC is due to the shrinking of the membrane [Tang *et al.*, (2008)].

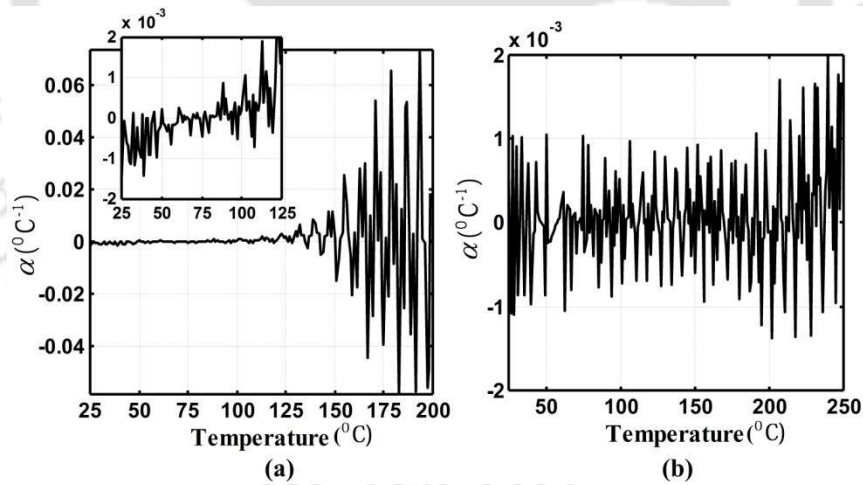


Figure 2.14 Variation of CTE (α) with temperature for (a) Nafion-117 film (b) pure Ag-IPMC membrane in a tensile mode.

Further, to investigate the effects of temperature and other influencing factors that may affect the linear thermal expansion mechanism of IPMC membrane, TMA analysis under tensile load is carried out in the same equipment. Data are collected at a heating rate of $5^{\circ}\text{C}/\text{min}$ up to 200°C for Nafion and 250°C for IPMC from room

temperature. The tensile load is adjusted to 0.1N with a frequency of 0.1Hz. A sample of 15 mm x 4 mm x 0.2 mm is prepared and tested on TMA and coefficient of thermal expansion, $CTE(\alpha)$ is obtained. The variation of α with temperature is shown in Figure 2.14. Figure 2.14(a) and 2.14(b) are showing the variation of coefficient of thermal expansion (CTE) of the Nafion film and Ag-IPMC membrane, respectively in a tensile mode. The CTE of Nafion-117 film (Figure 2.14(a)) is found to be around $0.002^{\circ}\text{C}^{-1}$ from room temperature to 125°C which is in good agreement with the previous result. As shown in Figure 2.14(b), it is observed that the CTE of Ag-IPMC shows a negative value from 25°C to around 75°C and then a positive value up to around 160°C with a maximum value of $0.0005^{\circ}\text{C}^{-1}$ and then decreases to zero at around 170°C . Between 170°C to 190°C the CTE value decreases and reaches a minimum value of $-0.00015^{\circ}\text{C}^{-1}$ beyond which the value further increases. The range between 25°C to 150°C is very important as the CTE gradually increases from negative to positive value and in between 150°C to 180°C the value gradually changes from positive to negative. This shows that the proposed Ag-IPMC has opposite thermal responses at different temperatures i.e. the IPMC exhibits bi-material property responding to temperature such that it shrinks when the temperature is lower than 75°C and scales up by sudden expansion when the temperature is beyond 75°C .

Interpreting the results demonstrated in Figure 2.13, it is observed that factors like experimental procedure i.e., compression/tensile mode, type of load i.e., constant/variable, and frequency of the applied load influences the linear thermal expansion characteristics of Nafion and IPMC membrane.

2.3.3 Mechanical Characteristics of IPMC Actuator

Several experiments are carried out to assess the bending characteristics, tip force, lifting capability, elastic modulus and damping characteristics and evaluated. Detailed experimental procedures and observations made are discussed in the following subsections.

2.3.3.1 Bending Characteristics

Mostly the fixed-free configuration of IPMC actuator is used for bending experiment because of its large attainable deflection. Figure 2.15 shows the experimental setup for determining the bending characteristics of an IPMC actuator of size $20 \times 5 \times 0.2 (\text{mm}^3)$ subjected to a DC input voltage. The actuator is fixed at one end where copper strips are used to provide the controlled voltage (0.2-1.2V) quasi-statically from a DC power source (0-32V, 0-2A). The other end of the actuator is free. During this process the bending of the IPMC is observed and video images are taken (Maker: Sony, Model: DSC-H20) which are then used to obtain the corresponding tip deflections.

It is observed that the IPMC reaches its equilibrium state around 15-20s after the voltage is applied. The power is switched off at around 25s. Considerable amount of time is given to settle any back relaxation effect and to reach the steady-state before tip deflection data $P(p_x, p_y)$ are measured. Figure 2.16 shows the tip position of the actuator with change in input voltage from 0.2V to 1.2V with an increment of 0.2V and the corresponding bending tip deflection data of the actuator for various input voltages is given in Table 2.4.

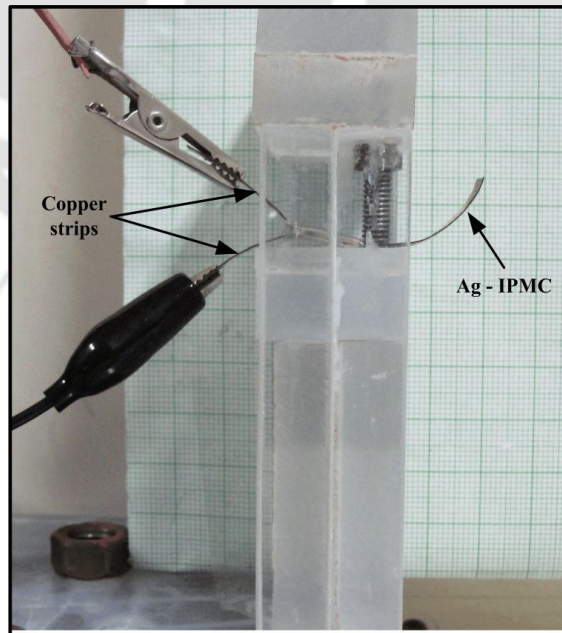


Figure 2.15 Actuation results of fabricated IPMC under DC electric potential of 1.2 V.

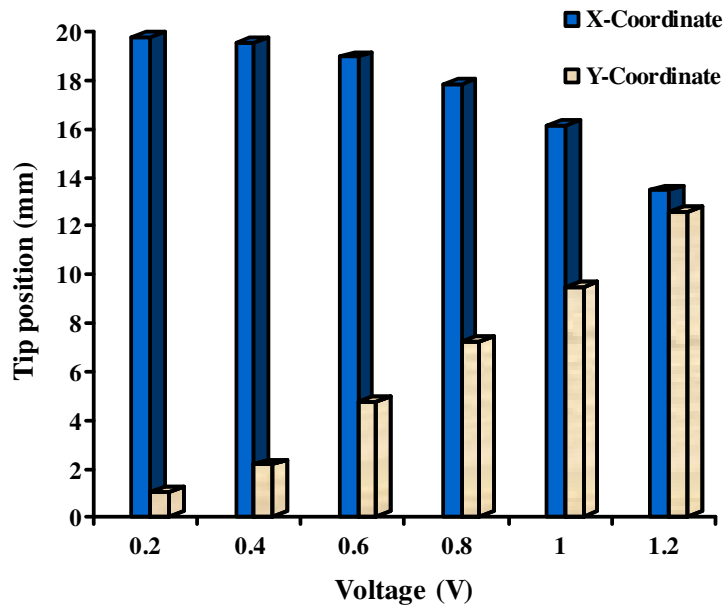


Figure 2.16 X and Y-Coordinates of the actuator with input voltage.

Table 2.4 Measured tip positions of the fabricated IPMC with various input voltage.

Input voltage (V)	0.2	0.4	0.6	0.8	1.0	1.2
Y-Coordinate (mm)	1.0	2.1	4.7	7.2	9.4	12.5
X-Coordinate (mm)	19.7	19.5	18.9	17.8	16.1	13.4

Figure 2.17 demonstrates the tip position of the IPMC actuator for various electrode materials such as IPMC actuator fabricated by solution casting method using silver nano-powder of size $20 \times 5 \times 0.2$ (mm^3), Chung *et al.*, (2006), gold plated IPMC of size $40 \times 5 \times 1$ (mm^3), Bandopadhyaya, (2007), platinum plated IPMC prepared by manual polishing of size $35 \times 7 \times 0.2$ (mm^3), Jin *et al.*, (2009) and IPMC fabricated by casting method with platinum as the surface electrode of size $50 \times 15 \times 0.45$ (mm^3), Peng *et al.*, (2009). It is observed that for an input of 1.0V, while the tip displacement of the present IPMC actuator is 9.4 mm, Chung *et al.*, (2006) reported 2.1 mm, Bandopadhyaya, (2007) reported 6.5 mm, Jin *et al.*, (2009) reported 7.1 mm, and Peng *et al.*, (2009) reported 2.31 mm, respectively. It is clearly observed that the

developed IPMC exhibits superior actuation performance compared to other surface electrode materials with less power consumption.

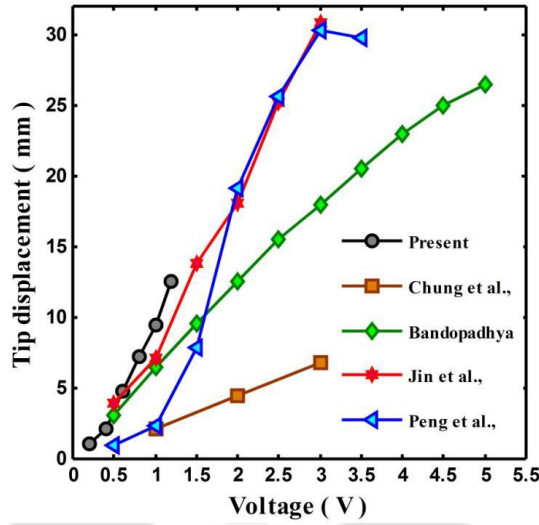


Figure 2.17 Comparison of tip displacement of various electrode materials with present work.

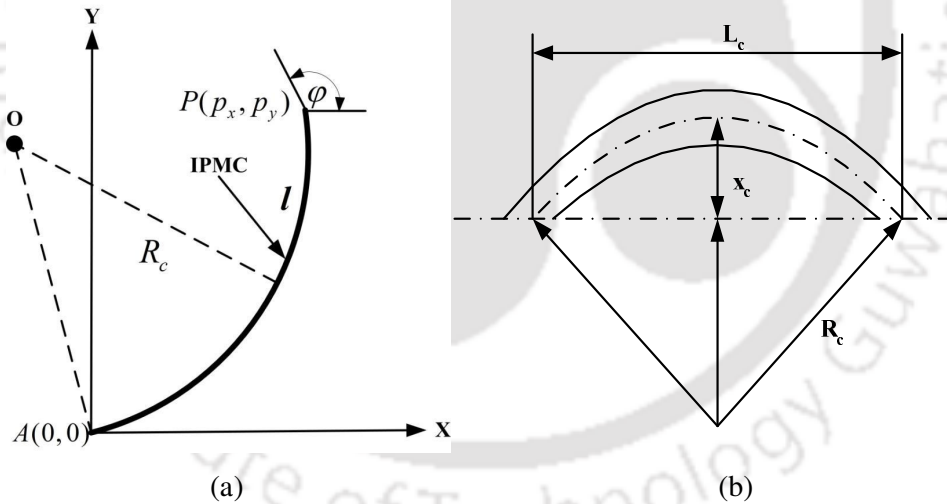


Figure 2.18 Schematic configuration showing bending of IPMC actuator under potential gradient (b) Schematic of radius of curvature of a curved IPMC.

Figure 2.18 shows the schematic diagram of the bending configuration of IPMC actuator under potential gradient. Assuming a constant curvature bending throughout the IPMC sample as observed during experiment (Figure 2.15) and the schematic diagram as shown in Figure 2.18 (a), the following relationship can be established:

$$\rho_c = \frac{1}{R_c} = \frac{\varphi}{l} \quad (2.2)$$

where, ρ_c is the bending curvature, R_c is the radius of curvature, l is the length of the IPMC, φ is the tip angle. However, for sake of validation, the experimentally obtained radius of curvature of the actuator is analyzed and it is found that the calculation based on tip angle and actual profile matches very well. Figure 2.19 (a) shows the relationship between voltage and tip angle obtained by using linear curve fitting approximation technique with a quality factor (R^2) 0.9729 and is given in equation (2.3).

$$\varphi = 1.4011V - 0.2924 \quad (2.3)$$

After measuring the tip deflection for each input voltage, equation (2.2) is used to obtain the voltage versus curvature relationship for both increasing and decreasing voltage and is plotted in Figure 2.19(b). It is observed that, the input-output relationship exhibits hysteresis.

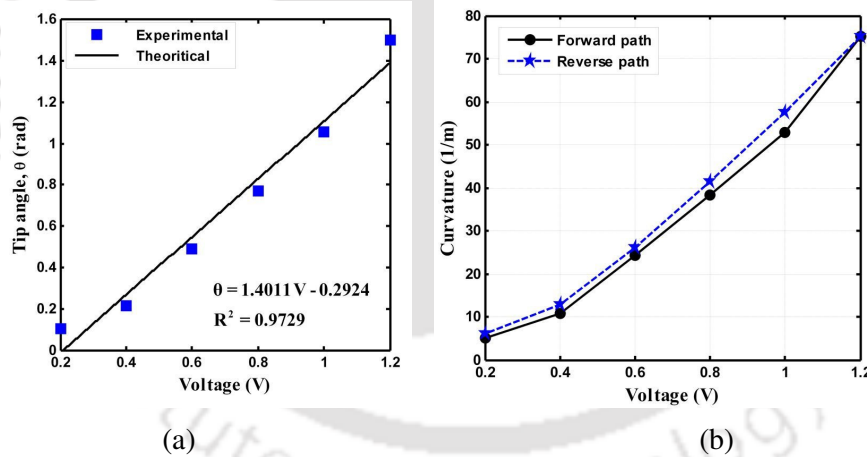


Figure 2.19 (a) Change in tip angle with input voltage (b) Voltage vs curvature relationship obtained experimentally.

2.3.3.2 Tip Force Measurement

Initially, before applying voltage at the fixed end of the actuator, the other end (free end) is just allowed to touch the digital measuring balance (Maker: Sartorius, Model: BSA2245-CW) and the initial reading is set to zero at time $t = 0$. An IPMC actuator of size $25 \times 6 \times 0.2 (\text{mm}^3)$ is experimented to measure the tip force using the setup as

shown in Figure 2.20(a). Voltage is applied gradually (not the step input) and the tip force is measured for an input of 1 V.

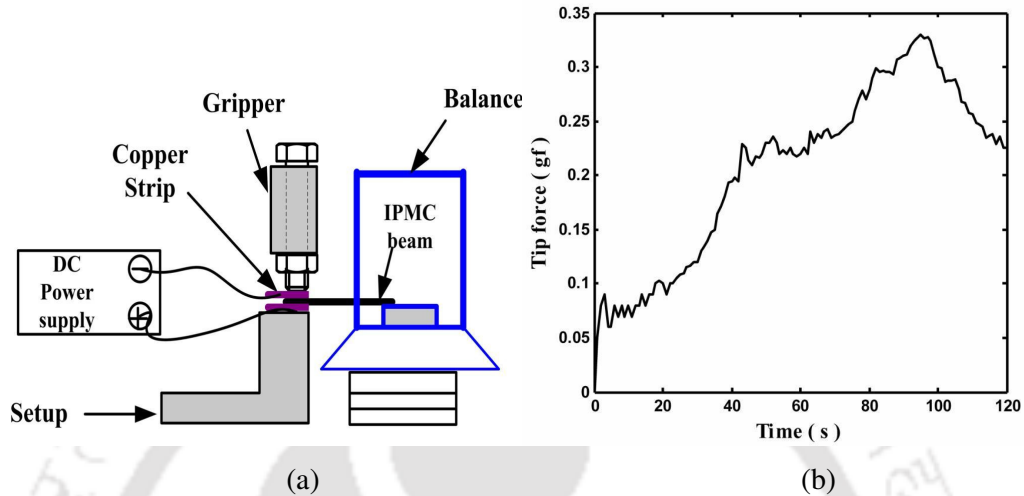


Figure 2.20 (a) Schematic diagram of the experimental setup for tip force measurement (b) Variation of generated tip force with time for a potential difference of 1.0V.

Figure 2.20(b) shows the force generated at the tip of the actuator for a time interval of 0 to 120s. The data are collected at an interval of 1s and is plotted. Tip Force generated depends on many factors like dimension (length, width and thickness), electrode materials, dehydration and applied voltage etc. Table 2.5 shows the comparison of the developed tip force for various electrode materials such as Ag, Pt, Au IPMCs with the present developed Ag-IPMC. From Table 2.5, it is observed that the developed IPMC generates much more force compared to other electrode materials such as the IPMCs of Ag, Pt, and Au etc.

Also the performance of the proposed IPMC is assessed by investigating lifting capacity i.e., force produced at the tip. Figure 2.21 shows the picture of the IPMC actuator of size $20 \times 5 \times 0.2$ (mm^3) at 3 different time intervals, which is used for finding the tip force/holding force of the actuator for an input of 0.6V. The weight of the payload is taken 0.58g, while the weight of the IPMC actuator is 0.0425g. It is observed that the fabricated IPMC can lift a weight of almost 14 times bigger than its own weight just for an input of 0.6V. Figure 2.22 demonstrates the holding force of the actuator at an input of 1.0V. Figure 2.22(a) shows the configuration of the same IPMC actuator with a payload of 2.1541g which is almost in the vertically

downward position. Figure 2.22(b) shows the position of the actuator at time $t = 3\text{s}$ for an input of 1.0V . The weight of the payload is taken 2.1541g , while the weight of the IPMC actuator is 0.0425g . It is observed that the fabricated IPMC can lift a weight of almost 50 times more than its own weight at this input of 1.0V .

Table 2.5 Comparison of developed tip force of various electrode material IPMCs.

Sl.No	Material and size	Tip force	Reference
1	Ag-IPMC [$20 \times 5 \times 0.23$ (mm^3), 1V]	0.114 gf	Chung <i>et al.</i> , (2006)
2	Pt-IPMC [$50 \times 15 \times 0.45$ (mm^3), 1V]	0.6667 gf	Peng <i>et al.</i> , (2009)
3	N-DIW, (Silver plated Nafion with water as solvent) [20×1 (mm^2), 1V]	0.2mN	Tamagawa <i>et al.</i> , (2003)
	N-H, (Silver plated Nafion with HCl as solvent) [20×1 (mm^2), 1V]	0.25mN	
	N-SC, (Silver, Copper plated Nafion with water as solvent) [20×1 (mm^2), 1V]	0.55mN	
	N-SCN, (Silver, Copper, Nickel plated Nafion with water as solvent) [20×1 (mm^2), 1V]	0.6mN	
4	Au-Ni-Nafion IPMC [11.5×4.7 (mm^2), 4V]	40mN	Siripong <i>et al.</i> , (2006)
5	Pt-IPMC [4×0.5 (cm^2), 3V]	1.5mN	Saher <i>et al.</i> , (2010)
6	Pt-IPMC [$40 \times 5 \times 0.4$ (mm^3), 1V]	3.34 mN	Lee <i>et al.</i> , (2011)
7	Ag-IPMC [$25 \times 5 \times 0.2$ (mm^3), 1V]	$0.33\text{gf} \approx 3.3\text{mN}$	Present work

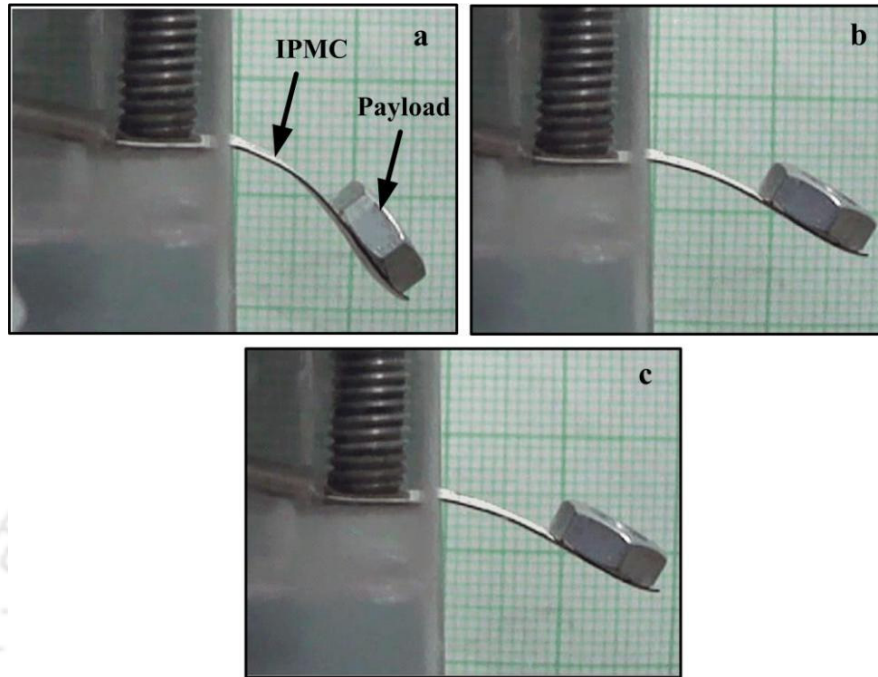


Figure 2.21 An Ag-IPMC lifting a payload with an input of 0.6V (a) $t=0s$, (b) $t=5s$, (c) $t=10s$

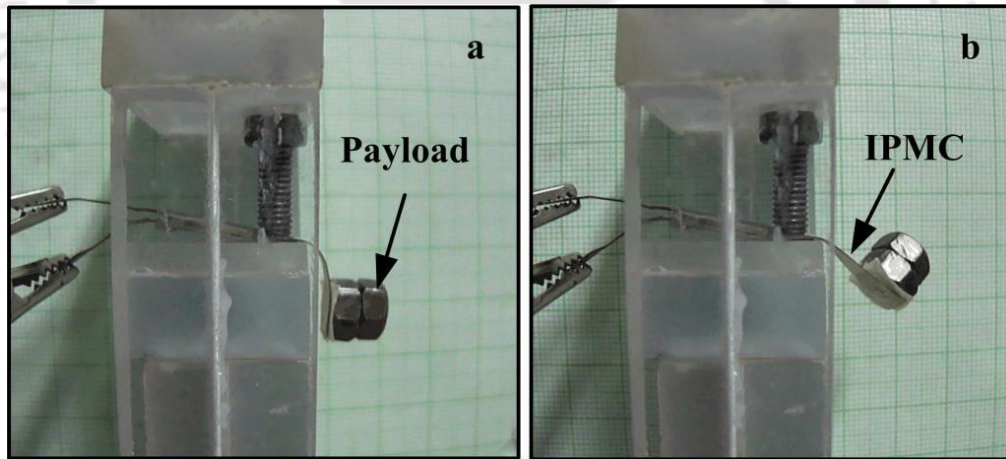


Figure 2.22 An Ag-IPMC lifting a payload with an input of 1.0V (a) $t=0s$, (b) $t=3s$

2.3.3.3 Evaluation of Elastic Modulus

Tensile tests are conducted to investigate the stress-strain behavior of Nafion and the IPMC, using a Micro-tensile-tester (Maker: Deben UK Ltd., Model: Microtest 5 kN). Figure 2.23(a) shows the micro tensile testing setup and the dimensions of the sample used for the test is shown in Figure 2.23(b). The tensile tests are performed

with a 0.5 mm/min test speed. Figure 2.24 shows the stress-strain curves obtained for both Nafion and IPMC in dry condition.

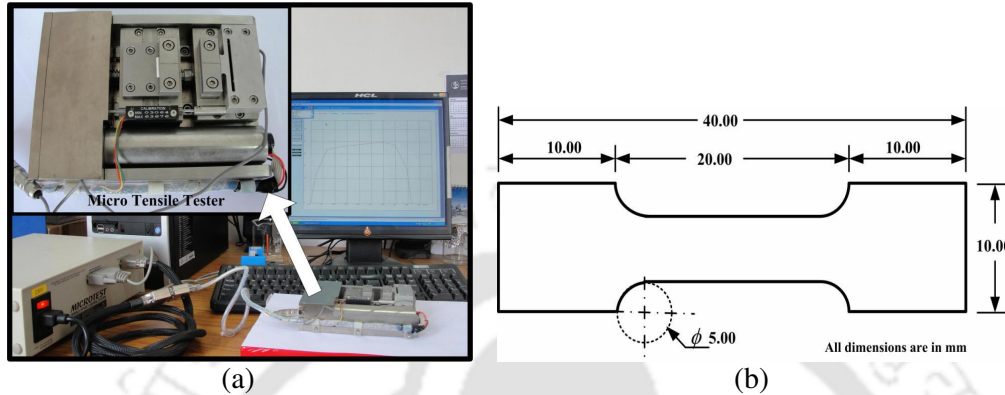


Figure 2.23 (a) Micro-tensile testing setup arrangement (b) Sample for micro-tensile test.

The Young's modulus of Nafion membrane is found to be 239 MPa, which is in good agreement with the results supplied by DuPont [DuPont™ PFSA membranes product information sheet]. The Young's modulus of Ag-IPMC in dry condition is found to be 422 MPa, which is higher than the base polymer Nafion due to presence of silver particles. Figure 2.25 shows the stress-strain curves obtained for IPMC in fully hydrated condition and the Young's modulus is found to be 82 MPa. It is observed that crack starts to initiate around 135s after the load is applied for Nafion and while for IPMC it is observed to be 250s. High Young's modulus of IPMC requires more electric potential to function properly and can not be used for many applications. Table 2.6 shows the comparison between various electrode-IPMC with the developed Ag-IPMC. The present result for wet Ag-IPMC is in good agreement with the results obtained by Tamagawa *et al.*, (2003) for N-DIW IPMC. Compared to the previously published results of Tamagawa *et al.*, (2003); Siripong *et al.*, (2006); Park *et al.*, (2007), the present results shows a moderate Young's modulus value, hence it is useful for application where low electric potential is required. Figure 2.26 shows the SEM micrograph of the IPMC after the tensile test. It is observed that, significant plastic deformation occurs and cracks are propagated before failure.

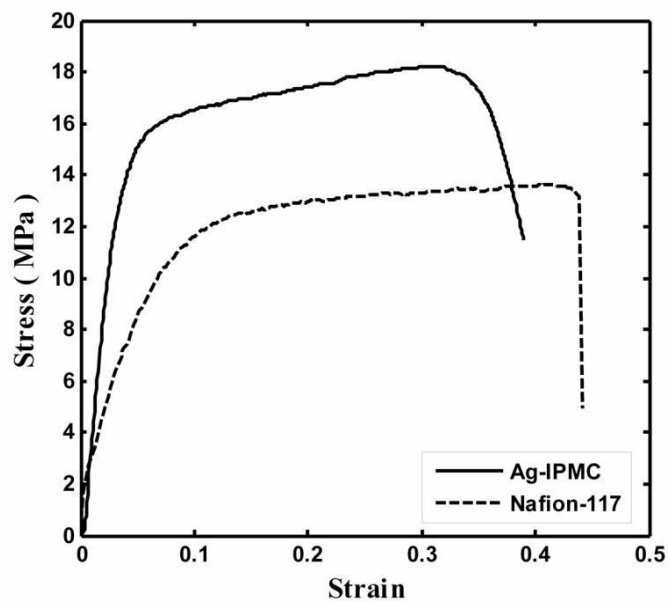


Figure 2.24 Stress-strain curves of Nafion and Ag-IPMC in dry condition.

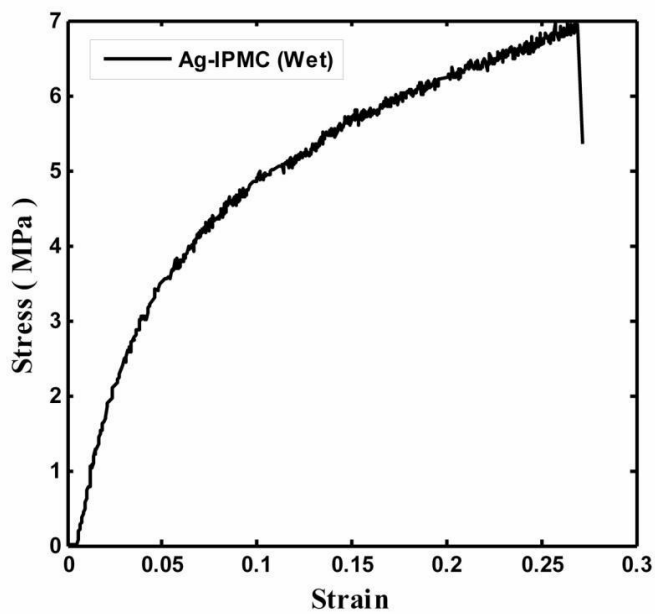


Figure 2.25 Stress-strain curves of Ag-IPMC in hydrated condition.

Table 2.6 Comparison of Young's modulus of various electrode material IPMCs.

Materials		Young's modulus (MPa)	Reference
Nafion™	Dried	131.6	Park <i>et al.</i> , (2007)
	Hydrated	30.4	
Pt-IPMC	Dried	114.3	
	Hydrated	41.2	
Au-IPMC	Dried	602.6	
	Hydrated	70.9	
Pd-IPMC	Dried	141.6	Tamagawa <i>et al.</i> , (2003)
	Hydrated	30.4	
IL-Pt-IPMC		20.6	
Ag-IPMC (N-DIW)	Swollen with deionised water	90	
Ag-IPMC (N-H)	Swollen with 0.1 M HCl	12	
Ag-Cu-IPMC (N-SC)		14	
Ag-Cu-Ni-IPMC (N-SCN)		124	Siripong <i>et al.</i> , (2006)
Nafion	Dried	245±9	
	Wet	139±31	
Ni-Au-IPMC	Dried	569±45	
	Wet	317±43	
Nafion	Dry	239	Present study
Ag-IPMC	Dry	422	
	Wet	82	

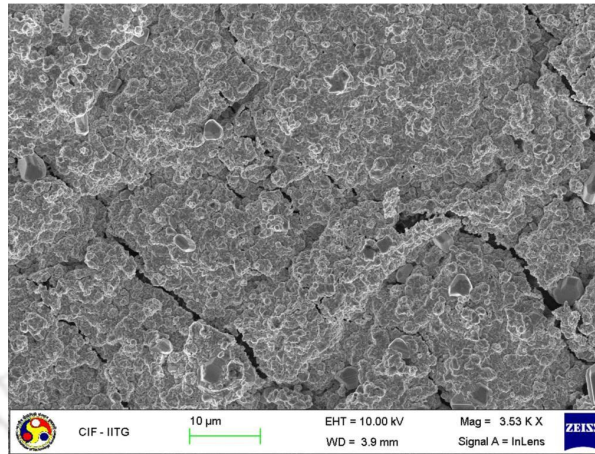


Figure 2.26 SEM micrograph of the IPMC after tensile test.

2.3.3.4 Damping Characteristics

The free vibration characteristic of an IPMC actuator of size $20 \times 5 \times 0.2 (\text{mm}^3)$ in a fixed-free mode is studied using Rotational Laser Vibrometer (RLV-5500), (Maker: Polytec, GmbH). The schematic diagram of the setup is given in Figure 2.27 and the photograph of the experimental setup is shown in Figure 2.28. The actuator is allowed to vibrate under a small deflection at the tip for a time span before it comes to rest. The vibration response is then measured using NI PXI-1031 controller with NI PXI-4472 module and LABVIEW software, supplied by National Instruments. The damping ratio and natural frequency of the actuator are investigated in fully hydrated (wet) condition. Figure 2.29(a) shows the free vibration response of the actuator.

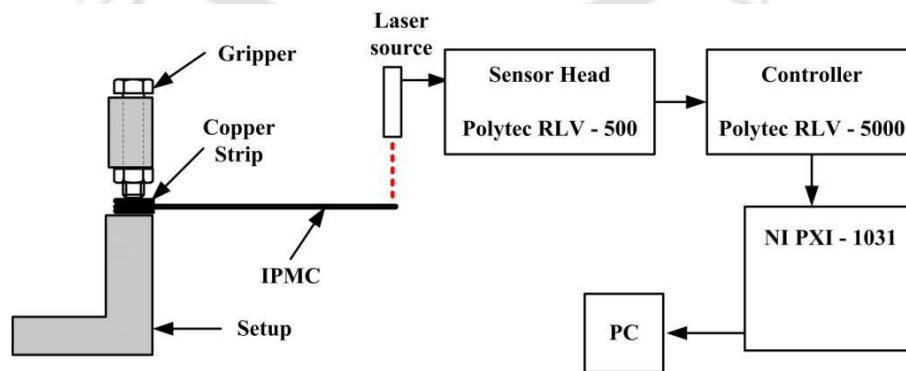


Figure 2.27 Schematic diagram of the free vibration setup.

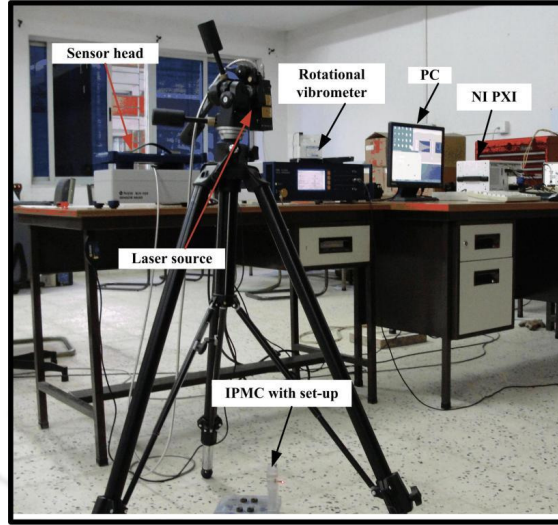


Figure 2.28 Photograph of the experimental setup for evaluation of damping characteristics.

Using logarithmic decrement method, the damping ratio of the IPMC is determined. The logarithmic decrement (δ) and the damping ratio (ζ) can be expressed as Craig, (1981):

$$\delta = \frac{1}{n} \ln \left(\frac{y_i}{y_{i+n}} \right) \quad (2.4)$$

$$\zeta = \sqrt{\frac{\delta^2}{4\pi^2 + \delta^2}} \quad (2.5)$$

where, n is the number of cycles, y_i is the amplitude of displacement of i^{th} cycle, y_{i+n} is the amplitude after n number of cycles. Taking ten successive cycles ($n=10$), the damping ratio (ζ) of the actuator is determined to be 0.018. Figure 2.29(b) shows the corresponding FFT of the vibration response. The quality factor (Q) is obtained from the forced vibration response and can be expressed as:

$$Q = \frac{1}{2\zeta} \quad (2.5a)$$

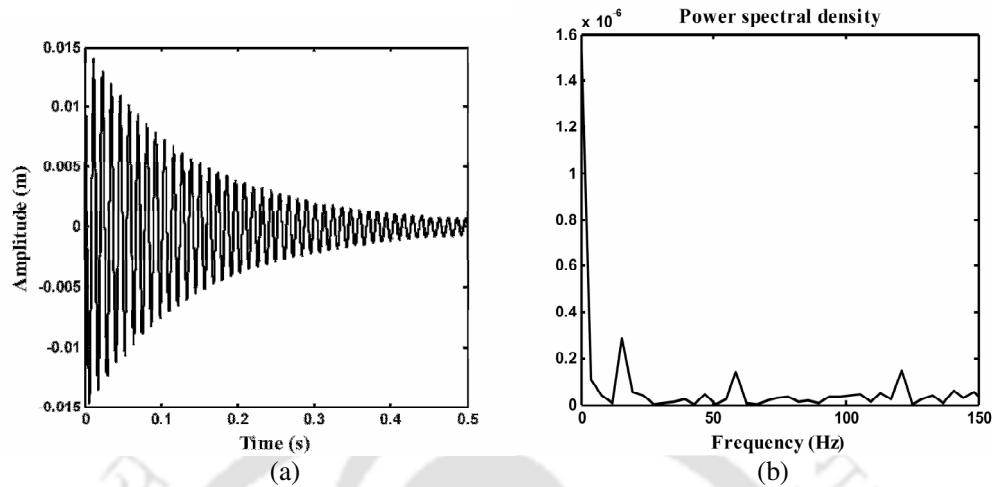


Figure 2.29 (a) Free vibration response (b) FFT at the tip of the actuator with a mechanical disturbance at the tip.

2.3.4 Surface Resistance of the Actuator

Surface resistance of the IPMC actuator is an important parameter influences the bending actuation and force generated by the actuator. The low surface resistance of IPMC improves the performance of the actuator [Shahinpoor and Kim, (2001a)] although the surface resistance changes due to bending deformation of the actuator [Punning *et al.*, (2007)].

In the present work surface resistance of IPMC is evaluated for both the cases i.e., static (without input voltage) and bending configuration (with input voltage). An IPMC actuator of size $40 \times 5 \times 0.2$ (mm³) is further studied in cantilever configuration subjected to a DC input voltage of 1.0V for measuring the change in surface resistance. Figure 2.30 shows the photograph of the bending sample after application of voltage. Surface resistance of Ag electrode layer is determined by using two-point AMPROBE 5XP-A multimeter. The data are collected between two points at a distance of 5 mm from base towards the tip of the sample and the average value over five measurements is taken and plotted as shown in Figure 2.31. It is observed that surface resistance gradually decreases over the length of the IPMC due to the bending deformation. This observation is in good agreement with the observation made by Punning *et al.*, (2007). The slight increase in the surface resistance at a

distance of 30mm as shown in Figure 2.31 may be due to the curvature effect of the bending part while taking the measurement.

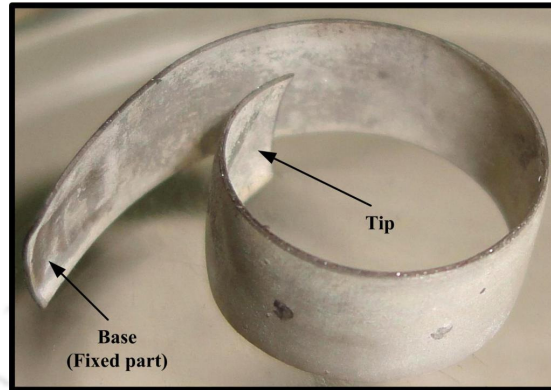


Figure 2.30 Photograph of the bending IPMC after application of voltage of 1.0V.

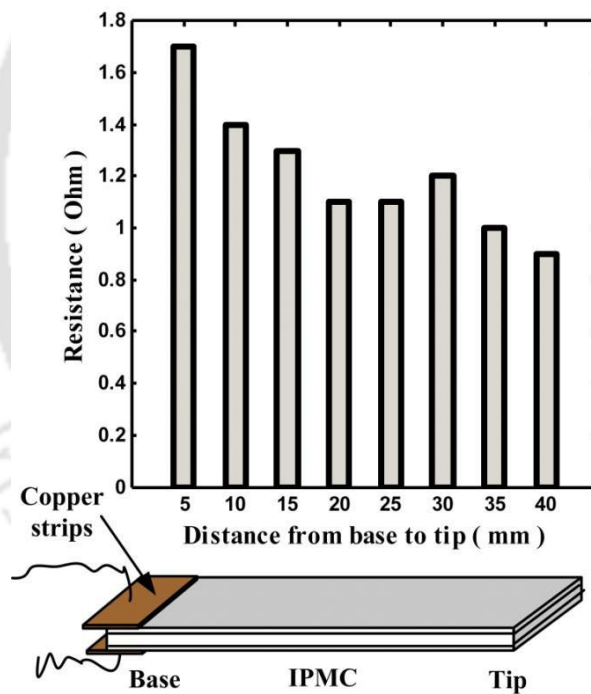


Figure 2.31 Surface resistance of IPMC sample with an input voltage of 1.0V.

Multiple measurements are also performed on a sample of same size in static configuration in wet condition at room temperature and the average value is obtained as 0.7 Ohm. Comparative studies for other IPMCs show the surface resistance as 1.7 Ohm [Chen *et al.*, (2009)], 3.84 Ohm [Shahinpoor and Kim,

(2001a)], 5.1 Ohm [Jin *et al.*, (2009)], 5.6 Ohm [Saher *et al.*, (2010)], 3.02 Ohm [Yip *et al.*, (2011)], and 6.15 Ohm [Lee *et al.*, (2006 b)] respectively. The study reveals that the developed IPMC is having low surface resistance than other IPMCs of electrode materials such as Pt and Ag.

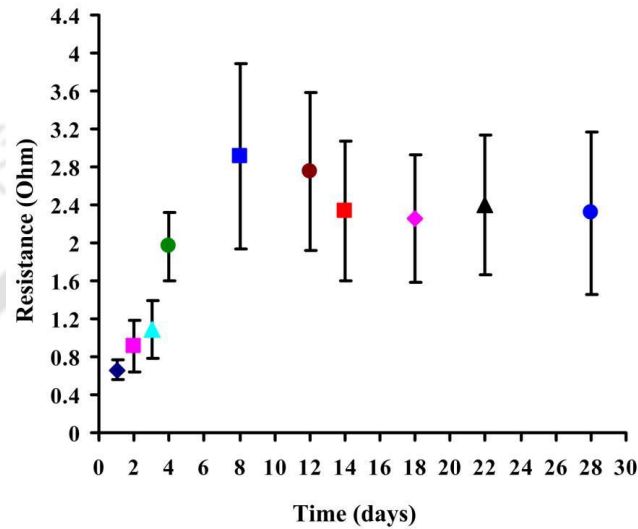


Figure 2.32 Relations between surface resistance and time without application of input voltage.

Figure 2.32 demonstrates the variation of surface resistance over time of the IPMC actuator. At the beginning, the surface resistance remains almost constant for 3 days, while it increases with time for about 8 days. Further increase in time, surface resistance remains to be stable less than 2.5 Ohm. The present result shows low surface resistance value compared to the Ag-IPMC developed by Chen *et al.*, (2009). From the result obtained by [Chen *et al.*, (2009); Figure 7(b)] it is observed that the surface resistance of Ag-IPMC increases gradually with time/days and observed to be less than 5 Ohm (surface resistance after day one around 1.7 Ohm, after 14 days around 3.74 Ohm, and after 16 days it is around 4.29 Ohm), while in the present case, surface resistance is measured as 2.4 Ohm after 14 days and remains stable around 2.5 Ohm. This implies that the method adopted for fabricating silver-electroded IPMC is better compared to the previous method. Hence, as evident, the low surface resistance implies better actuation and force generation capability of the actuator.

2.3.5 Solvent Loss Under Electric Potential

When an electrical potential is applied across the thickness of the IPMC actuator, the electromechanical transduction occurs due to transportation of cations with water molecules towards the cathode. Hydration level is one of the key factors which influence the bending deformation of the actuator. For conventional IPMC, water is used as the polar solvent. It is observed that gradual dehydration of moisture contents and electrolysis beyond 1.23V adversely affects the performance of the actuator. In the present study, several experiments are carried out to find out the solvent loss under DC and AC input voltages. Before the experiment, the samples are immersed in the distilled water, boiled for 30min and then kept for 24h. The weight of the samples before and after full hydration is measured to calculate the water uptake. Where, water uptake can be defined as:

$$M_{wu} = \frac{W_{fh} - W_{dry}}{W_{dry}} \quad (2.6)$$

where, M_{wu} is the water uptake, W_{fh} and W_{dry} is the weight of the fully hydrated and dried IPMC sample. Experimentally water uptake of an IPMC sample (20x5x0.2 (mm³)) is found to be 23 to 24%.

Solvent loss is calculated by experimenting a fully hydrated IPMC of size 20x5x0.2 (mm³). Both DC and AC (Sine wave, 1 Hz) voltage of 0.5V is applied for different time periods. The weight of the sample before and after the experiment (example: after applying input voltage for 10s, 20s etc.,) is measured using a digital measuring balance (Make: Sartorius, Model: BSA2245-CW). The solvent loss is defined as [Lee *et al.*, (2006 a)].

$$S_{loss} = \frac{W_i - W_f}{M_{wu}} \times 100\% \quad (2.7)$$

where, S_{loss} is the solvent loss, W_i is the weight of the IPMC before the voltage is applied and W_f is the weight measured after the voltage is applied for a certain time period.

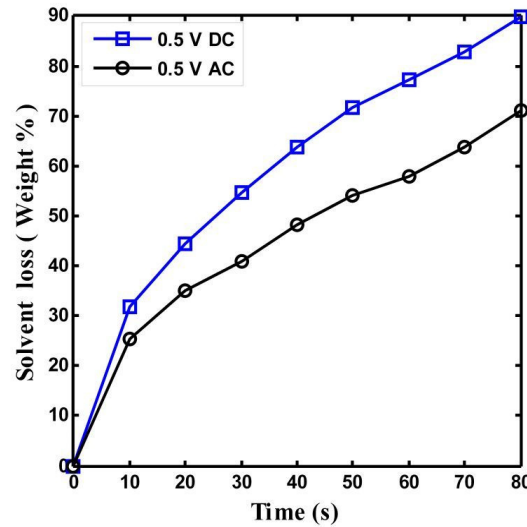


Figure 2.33 Solvent (water) losses from IPMC with input voltages.

Figure 2.33 shows the solvent loss from IPMC actuator for both AC and DC input voltage. It is observed that loss is much less when the actuator is operated with AC input compared to DC input signal. Further, it is observed that IPMC with 0.5 V DC input voltage can operate for a time period of 80s and suffered around 90% solvent loss. On the other hand, with 0.5V AC input, solvent loss is observed to be around 71%. The observation reveals that with AC input, the actuator can operate for a longer time period than same DC input voltage.

2.3.6 Operational Life Under DC Potential

IPMC works effectively in fully hydrated condition in open environment. However, during operation moisture dehydrates through the porous metal electrode significantly reduces the operational life of the actuator. Figure 2.5(a) demonstrates the thick dense homogeneous electrode layer of silver over the surface of the Nafion of the fabricated IPMC actuator. The electrode layer acts as water sealer to reduce the moisture loss from the IPMC.

An IPMC actuator of size $20 \times 5 \times 0.2$ (mm³) in fully hydrated condition is tested by applying 0.6 V DC to assess the operational life. The sample is tested for a time period of 10s, and then the polarity is reversed allowing the IPMC to deflect in opposite direction. The same experimental procedure is repeated until the IPMC

stops to exhibit bending deformation. Three samples have been tested, and it is observed that the average operating life of the IPMC is around 90-100s. The membrane can work again upon full hydration by putting it into deionized/distilled water. It is further observed that IPMC dehydrates more rapidly to DC actuation than AC.

2.3.7 Hydration Level and Temperature

The water uptake of an IPMC is related to the nano-structure of the membrane, the associated cation type, the temperature, and presence and concentration of the electrolyte [Gottesfeld and Zawodzinski, (1997)]. The effect of the temperature on the amount of water uptake is investigated using a fabricated IPMC. The weight of the IPMC sample is measured in dry state (W_{dry}). The sample is then immersed in the water having temperatures ranging from 30°C to 100°C and allowed to remain immersed for 1h to be soaked, after which, the weight of the wet IPMC samples ($W_{wet}(T)$) are measured. The hydration mass, M_h is defined as the mass of the absorbed water, ($W_{wet}(T) - W_{dry}$) divided by the mass of the sample in its dry state [Nemat-Nasser and Wu, (2006)]. The hydration mass M_h is calculated at each temperature. The water uptake (hydration mass) can be expressed as

$$M_h(T) = \frac{W_{wet}(T) - W_{dry}}{W_{dry}} \quad (2.8)$$

Figure 2.34 shows the change in water uptake with temperature. It is observed that, water uptake of IPMC increases linearly with the increase in hydration temperature. This observation is similar to the study of [Gebel *et al.*, (1993); Nemat-Nasser and Wu, (2006)].

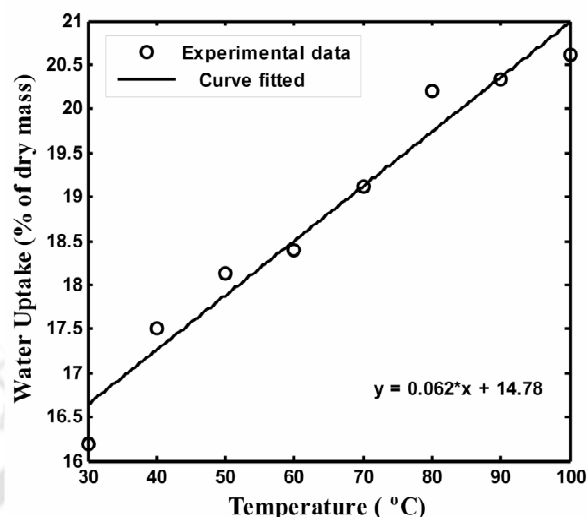


Figure 2.34 Water uptake of an IPMC at various temperatures.

2.4 Summary

The chemical decomposition method followed for depositing silver electrode layer over Nafion film appears to yield favorable results in the fabrication of Ionic Polymer-Metal Composites. Experimental study is carried out to characterize the electro-mechanical and thermal properties of Ag-IPMC. Topographical features are examined by scanning electron microscope suggested that there is a fair dispersion of silver particles over the polymer surface and also confirm adequate bonding between the metal and polymer layer. Also, SEM images show that silver particles penetrate through Nafion membrane up to 8-9 μm . Further, morphological studies of the fabricated IPMC show no crack formation in the micro-structure (with 1.17kX magnification). XRD analysis confirms that deposited silver film is polycrystalline in nature without any unwanted phase formation. In fact, it shows the characteristics of face-centred cubic (FCC) structure validates that silver-coated IPMC have perfect conductivity property.

DSC and TGA tests of the developed IPMC confirm improved thermal stability. The CTE of Nafion and IPMC films are investigated. It is concluded that the CTE of IPMC depends strongly on temperature. The CTE of IPMC shows negative value

with low temperature, and changes from negative to positive value with rise in temperature. Bending actuation results show a large bending deflection; as a result relatively large tip forces are achievable from these IPMCs using low input voltage and tensile test confirms the increase in Young's modulus. Measurement of surface resistance shows a comparatively low value than other surface electrode materials. As evident, the low surface resistance implies better actuation and force generation capability of the actuator. It is observed that the average operating life of the IPMC is around 90-100s. The membrane can work again upon full hydration by putting it into deionized/distilled water. The fabricated IPMC exhibits superior performance also shows no oxidation problem. Thus it is confirmed that the developed IPMC is inexpensive and suitable for application as an actuator and as well as artificial muscle materials.

Modeling and Evaluation of Dehydration Effect on Tip Position and Bending Response of IPMC Actuator

3.1 Introduction

The water content/hydration level of the base polymer of the IPMC serves as a medium for ion migration and thus influences the performance of the IPMC. Dehydration of moisture contents through the porous electrode surface reduces effectiveness of IPMC under working conditions. Further, for prolonged operation electrolysis occurs beyond certain input voltage along with dehydration due to applied electric potential results in decrease of the performance of IPMC.

Past literatures established the fact that the amount of water/moisture content is closely related to the bending performance of the IPMC actuator. However, very limited research has been carried out to address the dehydration problem. In this chapter, efforts are given to develop a simple but robust model from the experimental data to predict the loss of water content in terms of input voltage and time during working in the open environment. To estimate the moisture loss, experiment is conducted with an IPMC actuator with varying input voltage. At the starting of the experiment, the moisture content is assumed to be 100% and then gradually the loss of moisture/water content due to dehydration is estimated on account of loss in bending moment. Based on the experimentally obtained deflection data, a power fit correlation is established for loss-factor in terms of input voltage and time using Cobb-Douglas production method.

3.2 Calculation of Loss-Factor

A new term ‘loss-factor’ is introduced in this work for estimating the loss of moisture/water content from IPMC. Experiment is conducted with an IPMC actuator of size $20 \times 5 \times 0.2$ (mm³) (Figure 3.1) with varying input voltage. The IPMC used in the experiment is based on silver electrode with Nafion-117 as the base polymer (fabricated at Mechatronics laboratory, IIT Guwahati) as shown in Figure 3.1(a) while Figure 3.1(b) shows the experimental setup (which is same as Figure 2.15) to obtain the bending characteristics of the IPMC actuator subjected to input voltage (V).

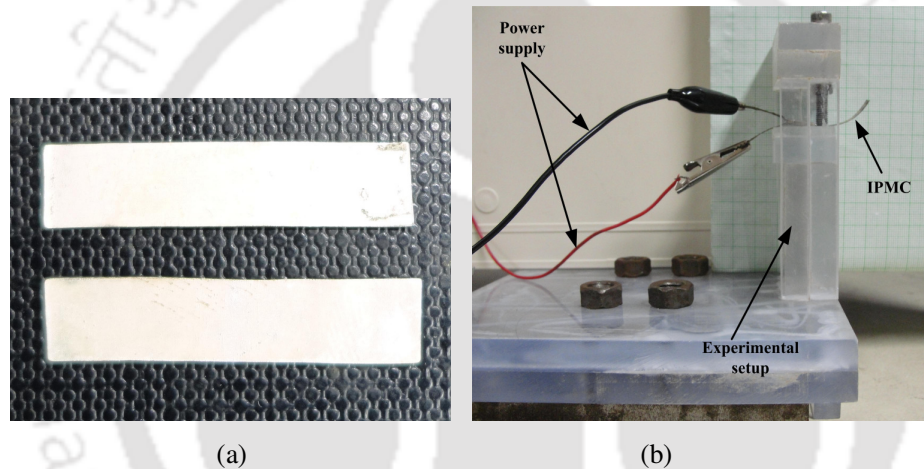


Figure 3.1 (a) Silver electroded IPMC fabricated at IIT Guwahati (b) Experimental setup.

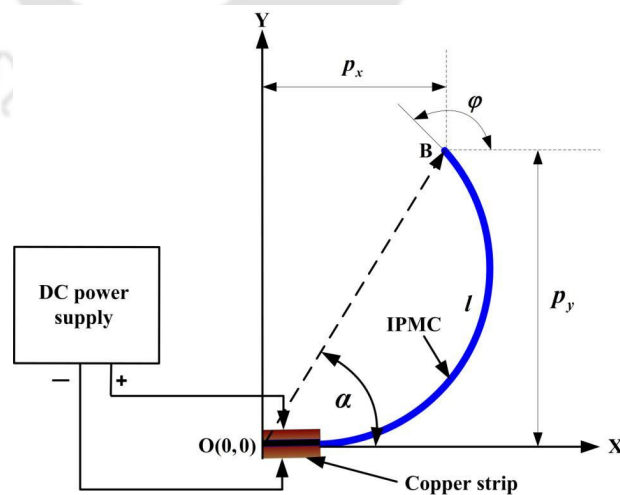


Figure 3.2 Schematic diagram of the experimental setup and bending configuration of an IPMC actuator for an input voltage V .

Figure 3.2 shows the schematic diagram for bending configuration. Experiment is conducted in hydrated state in fixed-free configuration by applying voltage quasi-statically from a DC power supply (0-32V DC, 0-2 A) through copper strips at the fixed end. Subsequently, the tip deflection of IPMC is measured and the bending characteristics are obtained. Experimentally dehydration loss is studied for the same IPMC sample subjected to input voltage from 0.2V to 1.2V with an increment of 0.2V. For each input voltage after 30s (sufficient time is allowed to settle any back relaxation effect and to attain the steady-state) the tip position (p_{xd}, p_{yd}) of the IPMC has been measured which are given in Table 3.1. The experiment is carried out for five samples and it is observed that the variation is negligible. The average value is given in Table 3.1.

Table 3.1 Tip position and angle of IPMC actuator with ‘positive’ and ‘zero’ dehydration.

Input voltage (V)	Time (s)	X- Coordinate (p_x) (mm)		Y-Coordinate (p_y) (mm)		Tip angle (rad)	
		p_{xd}	p_{x0}	p_{yd}	p_{y0}	ϕ_d	ϕ_0
0.2	30	19.7	19.8	1.0	1.1	0.10142	0.11098
0.4	60	19.5	19.6	2.1	2.3	0.21454	0.23362
0.6	90	18.9	19.0	4.7	5.1	0.48746	0.52446
0.8	120	17.8	18.0	7.2	8.0	0.76874	0.82714
1.0	150	16.1	16.0	9.4	10.8	1.0569	1.20238
1.2	180	13.4	12.8	12.5	14.5	1.50128	1.69516

Similar way, a fully hydrated IPMC sample has been taken for each time and experimentally the tip position (p_{x0}, p_{y0}) (zero dehydration), for each input voltage is obtained. After each experiment IPMC is retrieved and allowed to hydrate in aqueous deionized water and used for another set of experiment. At the starting of the experiment, the IPMC is considered fully hydrated i.e., moisture content is

considered 100% (reference value), while dehydration is zero. The obtained tip positions for ‘zero dehydration’ are given in Table 3.1. Following Bandopadhyaya *et al.*, (2009) the tip angle (φ) as shown in Figure 3.2 can be obtained as: $\varphi = 2 \tan^{-1}(p_y / p_x)$. The tip angle with ‘zero dehydration’ (φ_0) and ‘positive dehydration’ (φ_d) are calculated using the above formula and are given in Table 3.1 for various input voltages.

A term ‘loss factor’ (λ) is introduced by correlating loss in bending moment due to dehydration. As there is loss in bending moment due to dehydration during working, the bending moment at any instant (t) for an input voltage (V) can be expressed as:

$$M_d = M_0 - M_0 \lambda \quad (3.1)$$

where, M_d, M_0 are the bending moment of the actuator for ‘positive’ and ‘zero’ dehydration respectively.

Assuming pure bending, i.e., $M = \frac{EI}{R_c} = \frac{EI}{l/\varphi} = \frac{EI\varphi}{l}$, equation (3.1) reduces to

$$\frac{EI\varphi_d}{l} = \frac{EI\varphi_0}{l} (1 - \lambda) \quad (3.2)$$

Simplifying the equation (3.2), one can rewrite the equation as:

$$\lambda = 1 - \frac{\varphi_d}{\varphi_0} \quad (3.3)$$

where, EI is the flexural rigidity of IPMC and R_c is the radius of curvature; φ_d and φ_0 are the measurable parameters. As the IPMC dehydrates continuously in working condition, the primary goal is to develop an empirical model satisfying the working condition which can predict the loss of dehydration for any instant (t) and input voltage (V). Experimentally it is observed that the tip position and hence the tip angle depends on the applied voltage and actuation time while assuming temperature remains constant during the experiment. Therefore, the loss-factor also depends on these parameters and hence a power fit correlation is proposed as:

$$\lambda(V, t) = \gamma V^a t^b \quad (3.4)$$

where, Υ, a, b are the constants and depend on the material properties. Equation 3.4 also can be written as

$$\ln(\lambda(V, t)) = \ln(\Upsilon) + a \ln(V) + b \ln(t) \quad (3.5)$$

The goal is to develop a power fit correlation using Cobb-Douglas production method which correlates the loss-factor(λ) in terms of input voltage (V) and actuation time (t). Using equation 3.5 and the experimental data given in Table 3.1, a set of equations have been developed which are solved by the Cobb-Douglas production method. The resulting expression for the loss-factor is obtained as:

$$\lambda(V, t) = 0.9626 V^{0.7184} t^{-0.681}, \quad V \geq 0, t > 0 \quad (3.6)$$

Loss-factor for various voltages with time has been calculated subsequently and is given in Table 3.2. It is observed that with constant input voltage, loss-factor decreases continuously with time, although, it increases as the input voltage increases.

Table 3.2 Loss-factor for various input voltages and time.

Volt	Loss-factor (λ) with time (t)									
	10s	20s	30s	40s	50s	60s	70s	80s	90s	100s
0.2 V	0.0631	0.0394	0.0299	0.0246	0.0211	0.0186	0.0168	0.0153	0.0141	0.0132
0.4 V	0.1039	0.0648	0.0492	0.0404	0.0347	0.0307	0.0276	0.0252	0.0233	0.0217
0.6 V	0.1390	0.0867	0.0658	0.0541	0.0465	0.0410	0.0369	0.0337	0.0311	0.0290
0.8 V	0.1709	0.1066	0.0809	0.0665	0.0571	0.0505	0.0454	0.0415	0.0383	0.0356
1.0 V	0.2007	0.1252	0.0950	0.0781	0.0671	0.0592	0.0533	0.0487	0.0449	0.0418
1.2 V	0.2287	0.1427	0.1082	0.0890	0.0764	0.0675	0.0608	0.0555	0.0512	0.0477

Figures 3.3(a) and (b) show the variation of loss-factor with actuation time and applied voltage, respectively. These results are in good agreement with the observation that initial water content decreases with time, Enikov and Seo, (2005). Further, it is also observed that larger electric potential enhances more water loss Yeh and Shih, (2010). Further, it is observed that for a given input voltage, as the loss-factor decreases with time there is gradual reduction of actual bending moment and the tip deflection of the actuator. Table 3.3 shows both the estimated data (using equation (3.6)) and the observed data at zero dehydration for 30s.

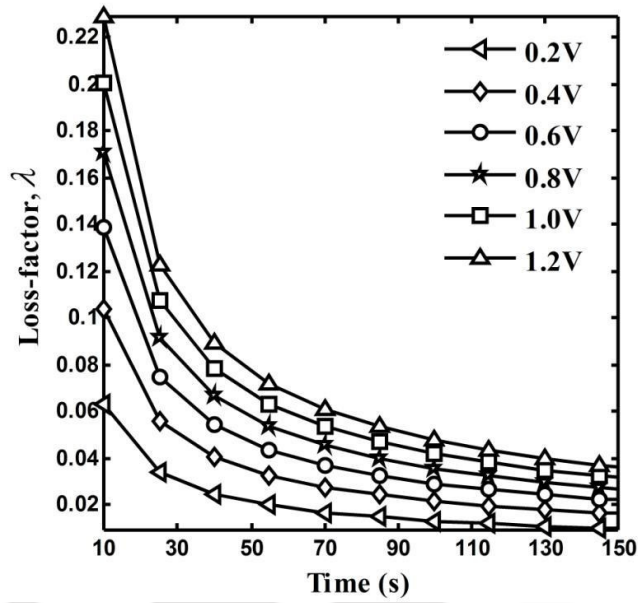


Figure 3.3 (a)

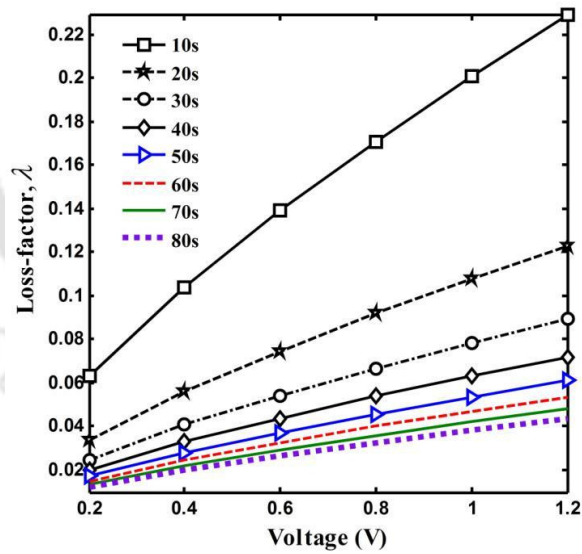


Figure 3.3 (b)

Figure 3.3 Variation of loss-factor with (a) time (b) input voltage.

Table 3.3 Comparison between the estimated data (using equation (3.3)) and the observed data (zero dehydration).

Voltage (V)	Observed data (mm)		Estimated data (mm)		% difference in X-Coordinate	% difference in Y-Coordinate
	p_x	p_y	p_x	p_y		
0.2	19.8	1.1	18.1096	1.0308	8.53	6.29
0.4	19.6	2.3	18.2748	2.2087	6.76	3.96
0.6	19.0	5.1	17.9479	5.0310	5.53	1.35
0.8	18.0	8.0	17.0362	7.8338	5.35	2.07
1.0	16.0	10.8	15.4272	10.3867	3.58	3.82
1.2	12.8	14.5	12.5206	14.0166	2.18	3.33

3.3 Bending Characteristics of IPMC

The experimental data obtained as given in Table 3.1 are used to establish relationship between bending moment and tip deflection with input voltage. Treating IPMC as continuous distributed parameter system with large deflection bending problem and using the Euler-Bernoulli bending-curvature relationship, one can establish the following relationship.

$$\frac{d\phi}{ds} = \frac{M}{EI} = \frac{d\phi}{dy} \frac{dy}{ds} = \frac{d\phi}{dy} \sin \phi \quad (3.7)$$

where, $\frac{d\phi}{ds}$ is the rate of change in angular deflection along the IPMC strip (curvature s), ϕ is the tip angle as shown in Figure 3.2, and EI is the flexural rigidity of IPMC. Separating the variables and integrating, tip position $P(p_x, p_y)$ of the actuator is obtained as:

$$\int_0^{p_y} dy = \frac{EI}{M} \int_0^{\phi} \sin \phi d\phi \quad (3.8)$$

$$p_y = \frac{EI}{M} (1 - \cos \phi) \quad (3.9)$$

$$\text{Similarly, } p_x = \frac{EI}{M} \sin \varphi \quad (3.10)$$

The relationship between bending moment and force generated by the actuator can be expressed as:

$$F = \frac{M}{p_x} \quad (3.11)$$

The experimental data for tip position (p_x, p_y) with an applied voltage (V) as given in Table 3.1 is plotted in Figure 3.4. Experimental results are approximated by 3rd order polynomial curve fitting technique and the following expression is obtained between (p_y) and (V).

$$p_y = -4.1667V^3 + 12.5V^2 + 1.1667V + 0.2, R^2 = 0.9971, 0.2 \leq V \leq 1.2 \quad (3.12)$$

Similarly,

$$p_x = -3.819V^3 + 0.476V^2 - 0.3948V + 18.8, R^2 = 0.9999, 0.2 \leq V \leq 1.2 \quad (3.13)$$

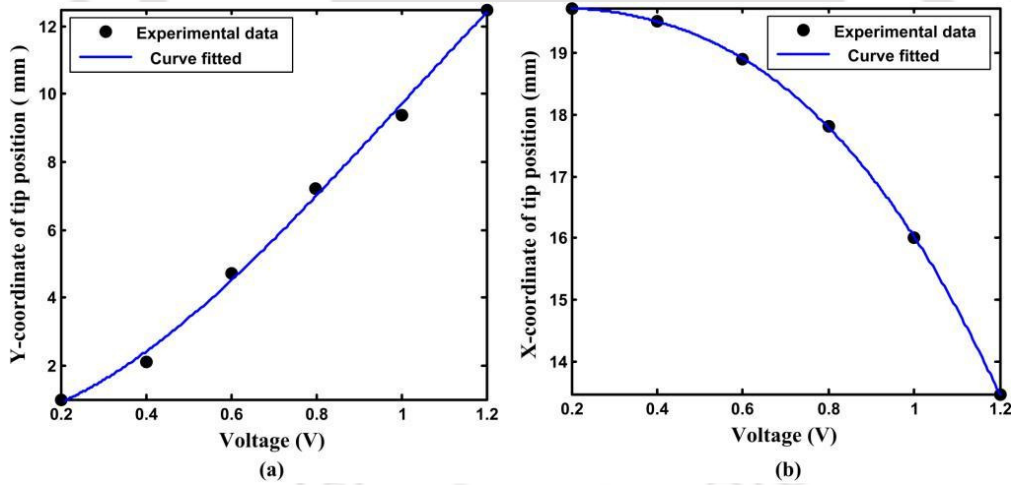


Figure 3.4 Change in tip position with input voltage (a) Y- coordinate (b) X- coordinate of the tip position.

Taking the material properties as given in Table 3.4, and using equation (3.9), the bending moment obtained for various input voltage is plotted and is shown in Figure 3.5. It is observed that the bending moment gradually decreases due to dehydration in working environment. The bending moment developed depends on the moisture

contents and subsequently the movement of ions within IPMC and the backbone materials.

Table 3.4 Physical properties and dimensions of IPMC actuator

Property	IPMC
E Elastic modulus	0.081877 GPa
l Length	0.02 m
w_d Width	0.005 m
h Thickness	0.0002 m
ρ Density	2125 kg/m ³
K Torsional spring constant	1.3646×10^{-5} Nm
m Mass	5.56×10^{-5} kg
C Damping Co-efficient	5.7712×10^{-8} Nm/rad

Figure 3.6 shows the effect of dehydration on the bending pattern of IPMC for various input voltages. Figure 3.7(a) and (b) shows the effect of dehydration on the tip position (X and Y-Coordinate) of the actuator, respectively. It is clearly observed from Figure 3.7(a), due to dehydration the tip position decreases.

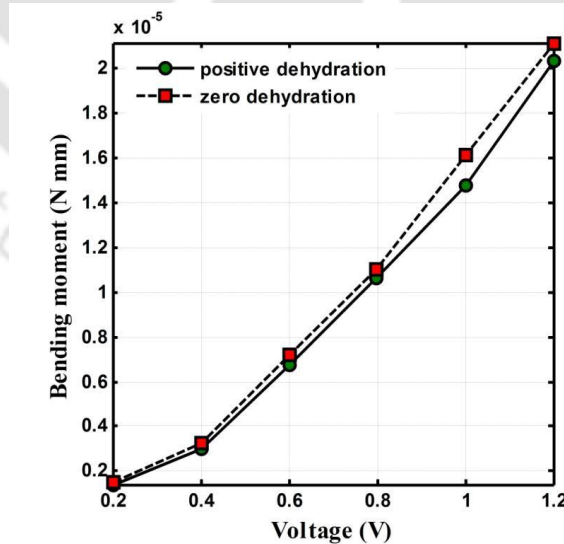


Figure 3.5 Bending moment generated for various input voltages.

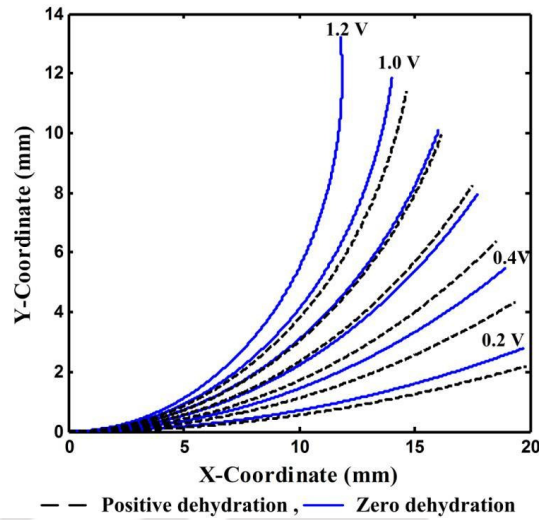


Figure 3.6 Bending pattern of IPMC for various input voltages.

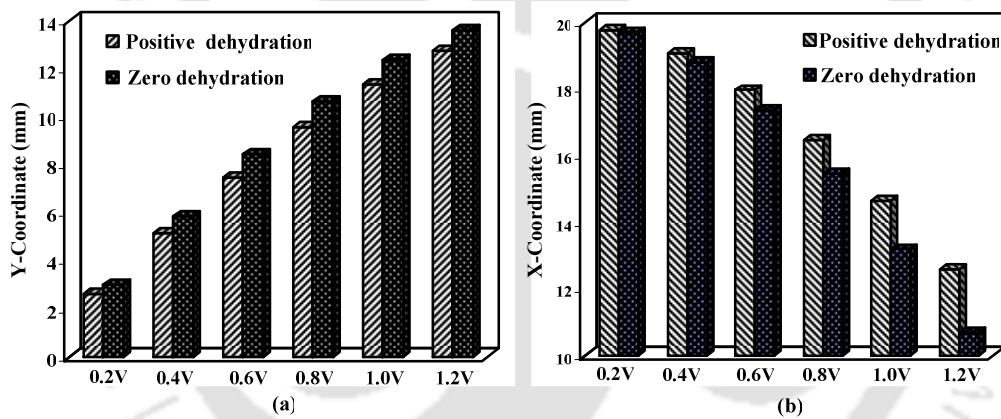


Figure 3.7 Effect of dehydration on the tip position of IPMC for various input voltages.

3.4 Modeling of Bending Characteristics

3.4.1 Euler-Bernoulli Approach

The experimental results obtained in the previous section are used to model the bending characteristics of IPMC following the Euler-Bernoulli approach. The maximum bending moment generated for each input voltage is given as: $M = \frac{EI}{R}$. An analogical assumption has been made that the bending phenomena is equivalent to the same amount of tip deflection caused by an external bending moment (M)

acting at the tip of the IPMC. This concept is employed to model the IPMC for each input voltage.

3.4.1.1 Single Patch IPMC Actuator

Applying Euler-Bernoulli approach, the tip position of the actuator for ‘positive’ and ‘zero’ dehydration, respectively can be obtained as, Bandopadhyaya (2009)

$$\begin{aligned} p_{xd} &= Q_d \sin \varphi_d \\ p_{yd} &= Q_d (1 - \cos \varphi_d) \end{aligned} \quad (3.14)$$

$$\begin{aligned} p_{x0} &= Q_0 \sin \varphi_0 \\ p_{y0} &= Q_0 (1 - \cos \varphi_0) \end{aligned} \quad (3.15)$$

where, $Q_d = \frac{EI}{M_0(1-\lambda)}$, $Q_0 = \frac{EI}{M_0}$

Using equations (3.14 and 3.15), the tip position of the IPMC actuator is plotted for various input voltages and is shown in Figure 3.8. While Figure 3.9 shows the tip positions for both positive and negative input voltages. It is observed that due to dehydration tip deflection of the IPMC actuator reduces considerably.

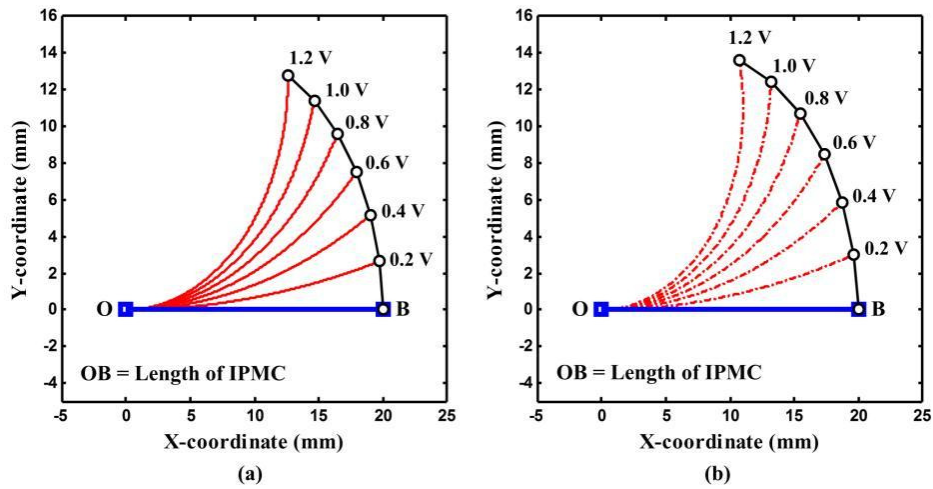


Figure 3.8 Tip position of the IPMC actuator (a) positive (b) zero dehydration following Euler-Bernoulli method.

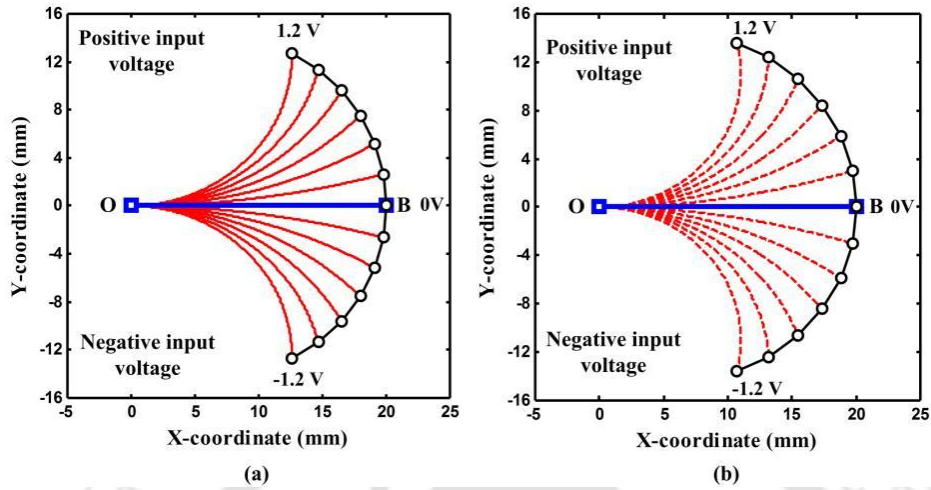


Figure 3.9 Tip position of the IPMC actuator for (a) positive (b) zero dehydration for both positive and negative input voltage following Euler-Bernoulli method.

3.4.1.2 Segmented IPMC Actuators

The attribute of developing bending deflection towards anode under potential gradient and ease of miniaturization makes IPMC actuator suitable for micro-scale manipulation. It is anticipated that, by segmenting IPMC strip in many parts (patches) and applying different sequence of voltage on different strips, enables one to control the deflection of end-tip in a manner as shown in Figure 3.10. The concept allows one to fabricate and design IPMC for manipulating micro-objects on a surface. Obviously, the work volume depends on the length of the strip and the number of parts of a strip for a desired position of the end point.

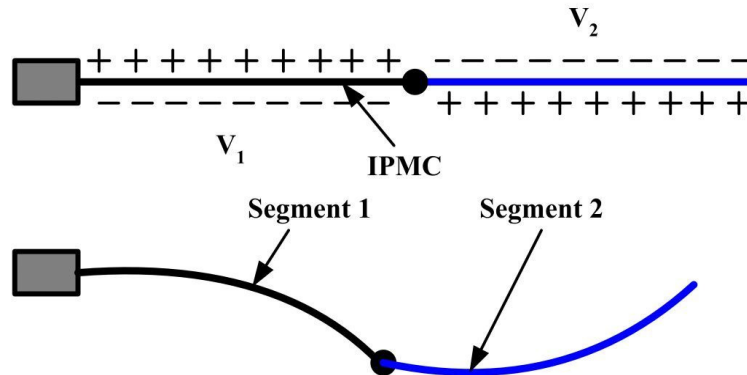


Figure 3.10 Bending configurations of two segmented patches of IPMC for both positive and negative sequence of input voltage.

The response of the segmented patches depends on the orientation and the applied voltage sequence. Each segment of IPMC may be subjected to different sequence of input voltage and in a way, the tip of each segment follow a path on a surface. Homogeneous coordinate transformation matrix is constituted based on the tip position with respect to the base coordinate. The generalized transformation matrix in 2D is given by:

$$A_{i+1}^i = \begin{bmatrix} \cos \varphi_i & -\sin \varphi_i & p_{xi} \\ \sin \varphi_i & \cos \varphi_i & p_{yi} \\ 0 & 0 & 1 \end{bmatrix} \quad (3.16)$$

where, (p_{xi}, p_{yi}) are the position co-ordinates. For the 1st and 2nd patch the transformation matrix is given by:

$$A_1^0 = \begin{bmatrix} \cos \varphi_1 & -\sin \varphi_1 & Q \sin \varphi_1 \\ \sin \varphi_1 & \cos \varphi_1 & Q(1 - \cos \varphi_1) \\ 0 & 0 & 1 \end{bmatrix} \quad (3.17)$$

$$A_2^1 = \begin{bmatrix} \cos \varphi_2 & -\sin \varphi_2 & Q \sin \varphi_2 \\ \sin \varphi_2 & \cos \varphi_2 & Q(1 - \cos \varphi_2) \\ 0 & 0 & 1 \end{bmatrix} \quad (3.18)$$

where, (φ_1, φ_2) are the orientations of patch 1 and 2 respectively. Therefore, with respect to base,

$$A_2^0 = A_1^0 \cdot A_2^1 = \begin{bmatrix} \cos(\varphi_2 + \varphi_1) & -\sin(\varphi_2 + \varphi_1) & Q \sin(\varphi_2 + \varphi_1) \\ \sin(\varphi_2 + \varphi_1) & \cos(\varphi_2 + \varphi_1) & Q(1 - \cos(\varphi_2 + \varphi_1)) \\ 0 & 0 & 1 \end{bmatrix} \quad (3.19)$$

Thus, tip position is obtained as:

$$\begin{aligned} p_{x_2} &= Q \sin(\varphi_2 + \varphi_1) \\ p_{y_2} &= Q(1 - \cos(\varphi_2 + \varphi_1)) \end{aligned} \quad (3.20)$$

Figure 3.11(a) and (b) are showing the tip position for segmented IPMC actuators for the same sequence of input voltage with dehydration. Figure 3.12(a) and (b) demonstrate the effect of dehydration on the tip position of the IPMC actuator for various input voltages. It is observed that due to dehydration tip deflection of the IPMC actuator reduces considerably.

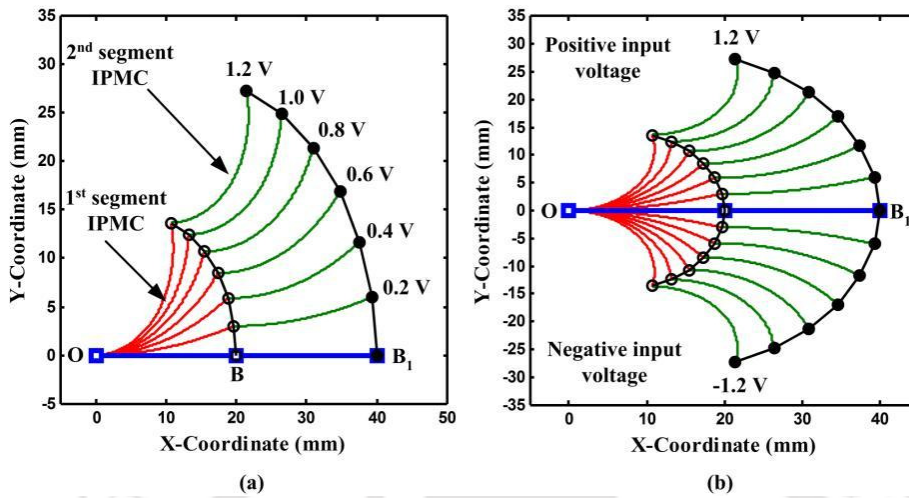


Figure 3.11 Bending of two patches (a) positive (b) both positive and negative sequence of input voltages following Euler-Bernoulli model.

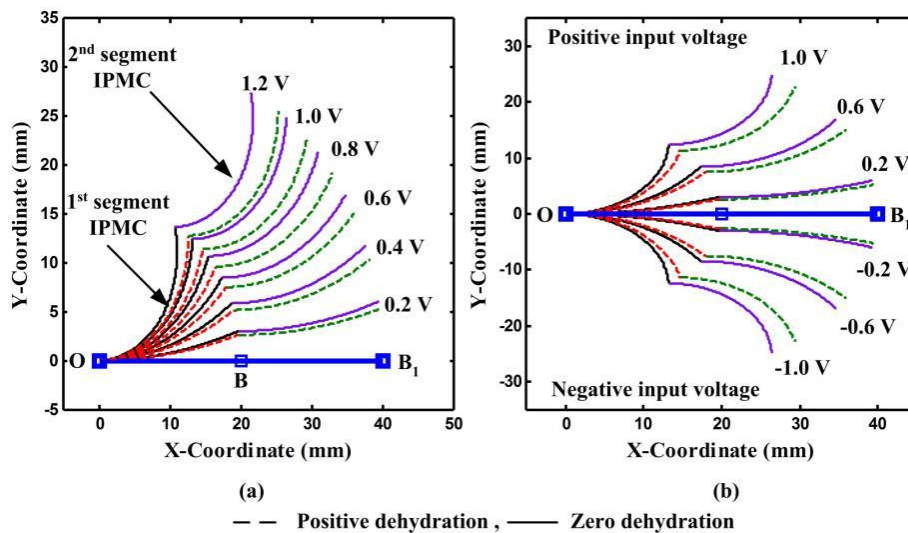


Figure 3.12 Effect of dehydration on segmented IPMC actuator (a) positive (b) both positive and negative sequence of input voltages following Euler-Bernoulli model.

3.4.2 Pseudo Rigid Body Modeling Approach

The pseudo-rigid body modelling technique is employed to simplify the analysis of compliant mechanism. The model approximates the IPMC actuator a compliant member and converts it into an appropriate equivalent rigid-body mechanism. The model is based on the fixed-free type mode with end-moment loading. The correlation function between the bending moment (M) and the spring constant (K)

with the other kinematic parameters such as characteristics radius factor (γ) and the pseudo-rigid body angle (θ) with applied voltage has been derived using the proposed technique. A generalized model is developed for segmented IPMC actuator treating it as a serial-link manipulator. It is assumed that each link bends with a uniform curvature with constant modulus of elasticity and the motion of the manipulator is restricted to two dimensional spaces.

In the present work an IPMC actuator has been studied to assess the effect of loss due to dehydration on bending resistance and bending response during working condition. Variable parameter pseudo-rigid body modelling (VPPRBM) technique has been used to derive the forward bending model of the actuator. The motivation is to assess the change in bending resistance of IPMC actuator with dehydration by applying pseudo-rigid body modelling technique.

3.4.2.1 Single Patch IPMC Actuator

Fixed Parameter Pseudo-Rigid Body Modeling

Pseudo-rigid body modeling of IPMC is dominated by the bending moment generated due to the applied voltage. Bending moment generated in IPMC can be expressed by $M = K\theta$ where, M is the bending moment, K is the torsional spring constant, and θ is the corresponding pseudo-rigid body angle [Bandopadhyaya, (2009)]. The spring constant is directly obtained as for the model as shown in Figure 3.15 [Howell, (2001)]

$$K = 1.5164 \frac{EI}{l} \quad (3.21)$$

End point coordinates of the link as shown in Figure 3.13 at dehydrated condition (experimented results) can be obtained as [Bandopadhyaya, (2009)]

$$p_x = l(1 - \gamma) + \gamma l \cos \theta \quad (3.22)$$

$$p_y = \gamma l \sin \theta \quad (3.23)$$

where, γ is the characteristic radius factor.

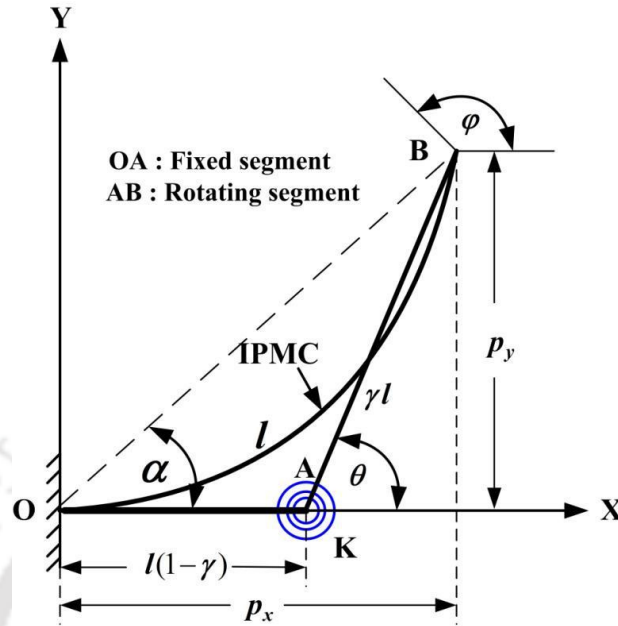


Figure 3.13 Pseudo-rigid body models for the deflected IPMC.

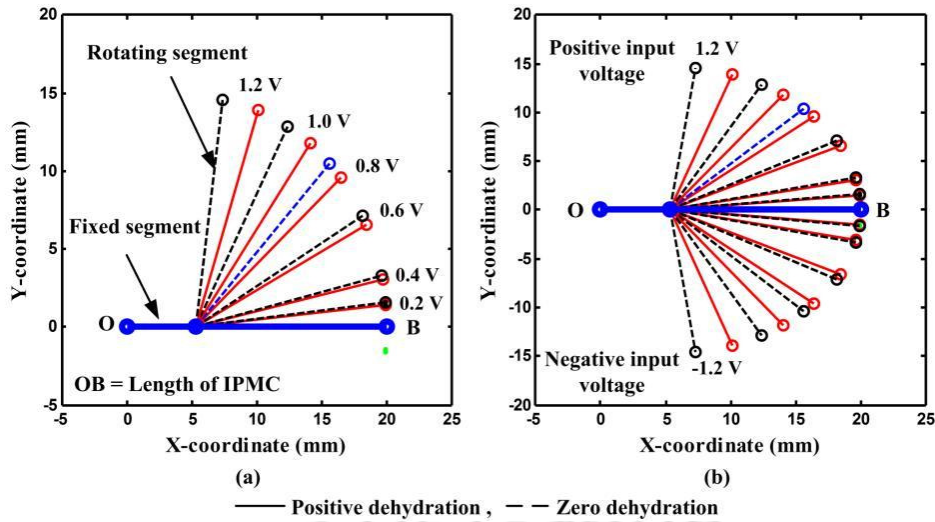


Figure 3.14 Change in tip position of the IPMC actuator following fixed pseudo-rigid body modelling for (a) positive (b) both positive and negative input voltage.

Figure 3.14(a) and (b) demonstrate the effect of dehydration on the tip position of the IPMC actuator for various input voltages using fixed parameter pseudo-rigid body modeling technique.

Variable Parameters Pseudo-Rigid Body Modeling (VPPRBM)

The experimental results obtained in the previous section are used to model the bending characteristic of the IPMC following the variable parameter pseudo-rigid body modelling (VPPRBM) technique. The technique validates the material properties and assesses the bending resistance offered by the IPMC during actuation, [Howell, (2001)].

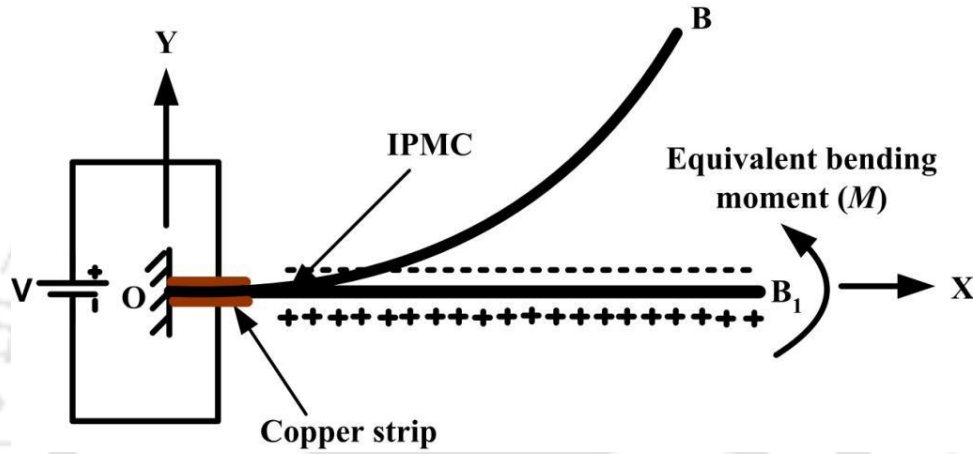


Figure 3.15 Equivalent bending model of IPMC for an input voltage V .

Figure 3.15 shows the initial (OB_1) and bending (OB) configuration of the actuator. Figure 3.13 shows the dual schematic representation of the bending configuration of IPMC with its pseudo-rigid body model. The IPMC is replaced by a link with a characteristic pivot located at a length γ_l from the free end. A torsional spring at the pivot measures the resistance to its bending. Bending moment developed by the actuator can be expressed as $M_d = K_v \theta_v$ where, K_v, θ_v are the variable spring constant and variable pseudo-rigid body angle respectively. The main objective of the analysis is to establish relationships among the parameters such as variable spring constant (K_v), variable pseudo-rigid body angle (θ_v) and variable characteristic radius factor (γ_v) with positive dehydration. End point coordinates of the link as shown in Figure 3.13 at dehydrated condition (experimented results) can be obtained as [Bandopadhyaya and Njuguna (2009)]:

$$p_x = l(1 - \gamma_v) + \gamma_v l \cos \theta_v \quad (3.24)$$

$$p_y = \gamma_v l \sin \theta_v \quad (3.25)$$

where, p_x, p_y are the end point coordinates at dehydrated condition.

The product $\gamma_v l$ (AB), the variable characteristic radius, is the radius of the circular deflection path traversed by the end-tip of the pseudo-rigid body link and $l(1 - \gamma_v)$ represents the length of the characteristic pivot located from the fixed end 'O' [Bandopadhyaya and Njuguna (2009)]. Further, as shown in Figure 3.13

$$OA + AB = l \Rightarrow l(1 - \gamma_v) + \sqrt{[p_x - l(1 - \gamma_v)]^2 + p_y^2} = l \quad (3.26)$$

Simplifying the equation (3.26)

$$l(1 - \gamma_v) = \frac{(p_x^2 + p_y^2 - l^2)}{2(p_x - l)} \quad (3.27)$$

Therefore, one can establish the relationship for characteristic radius factor as:

$$\gamma_v = 1 - \left[\frac{(p_x^2 + p_y^2 - l^2)}{2l(p_x - l)} \right] \quad (3.28)$$

Subsequently, pseudo-rigid body angle (θ_v) is obtained as:

$$\theta_v = \sin^{-1} \left(\frac{p_y}{AB} \right) = \sin^{-1} \left(\frac{p_y}{\gamma_v l} \right) \quad (3.29)$$

Treating IPMC as continuous distributed parameter system with large deflection bending problem, the bending moment generated by the actuator is obtained as,

$$M_v = \frac{EI}{p_y} (1 - \cos \varphi), \text{ (equation (3.9))}, \text{ and subsequently in terms of pseudo-rigid}$$

body angle can be expressed as:

$$M_v = \frac{EI(1 - \cos \varphi)}{\gamma_v l \sin \theta_v} \quad (3.30)$$

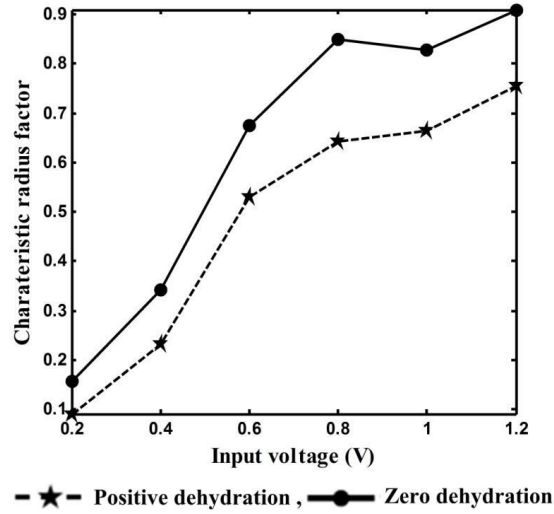


Figure 3.16 Effect of dehydration on the characteristic radius factor of IPMC with various input voltages.

The variable spring constant for the IPMC in dehydrated condition can be expressed as:

$$K_v = \frac{M_v}{\theta_v} \quad (3.31)$$

Figure 3.16 shows the variation of characteristics radius factor for both hydrated and dehydrated condition, while Figure 3.17 shows the change in the variable pseudo-rigid body angles. It is observed that with the input voltage, the variable pseudo-rigid body angle increases with dehydration and comes closer to a value of hydrated condition with increase in voltage. It is anticipated that the percentage differences in y-deflection between hydrated and dehydrated condition increases with input voltage while the characteristics radius factor decreases gradually. As the pseudo-rigid body angle depends both on y-deflection and the radius factor as given in equation (3.30), thus at low operating voltage, pseudo-rigid body angle increases with dehydration compared to hydrated state as shown in Figure 3.17.

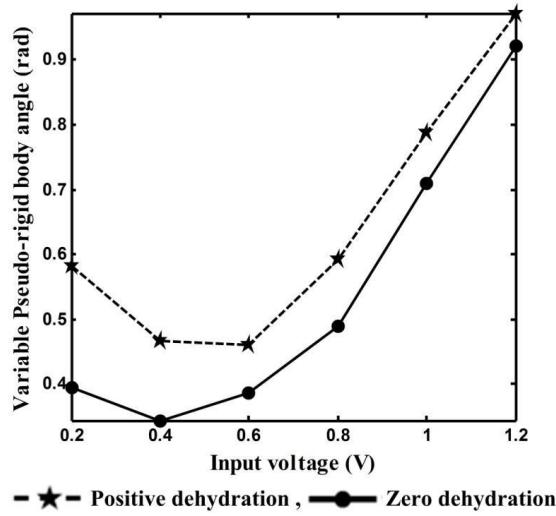


Figure 3.17 Change in pseudo-rigid body angle due to dehydration for various input voltage.

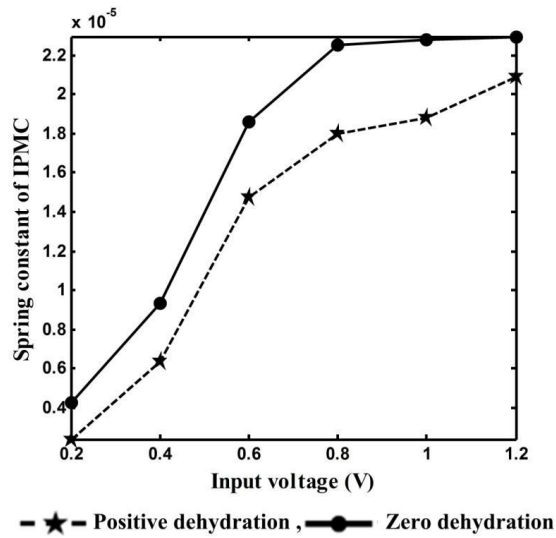


Figure 3.18 Variation of spring constant of IPMC with various input voltages.

Figure 3.18 is showing the change in spring constant with input voltage. The spring constant measures the resistance offered by IPMC during bending. It is observed that both spring constant and bending moment decreases with dehydration and comes to a closer value after certain time interval. Thus the proposed model validates the results pertaining to the assumptions made and the material properties taken.

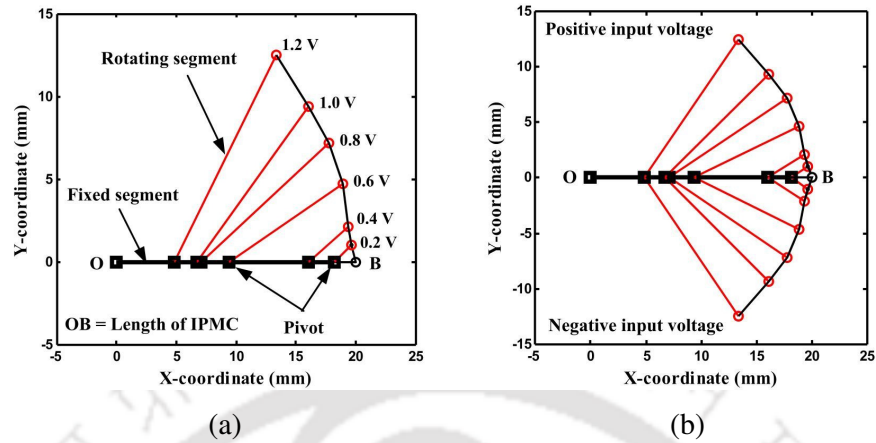


Figure 3.19 Path followed by the IPMC actuator using variable parameter pseudo-rigid body modelling for (a) positive (b) both positive and negative input voltages.

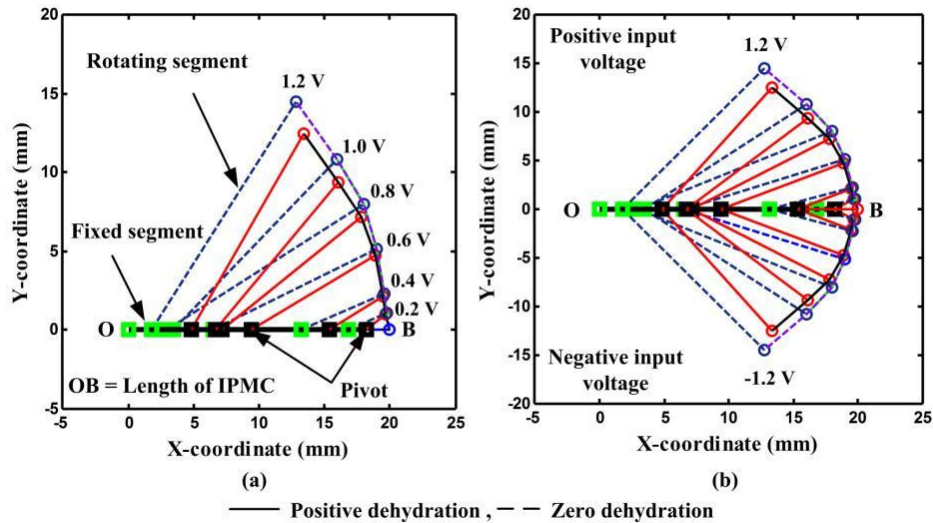


Figure 3.20 Effect of dehydration following variable parameter pseudo-rigid body modelling for (a) positive (b) both positive and negative input voltages.

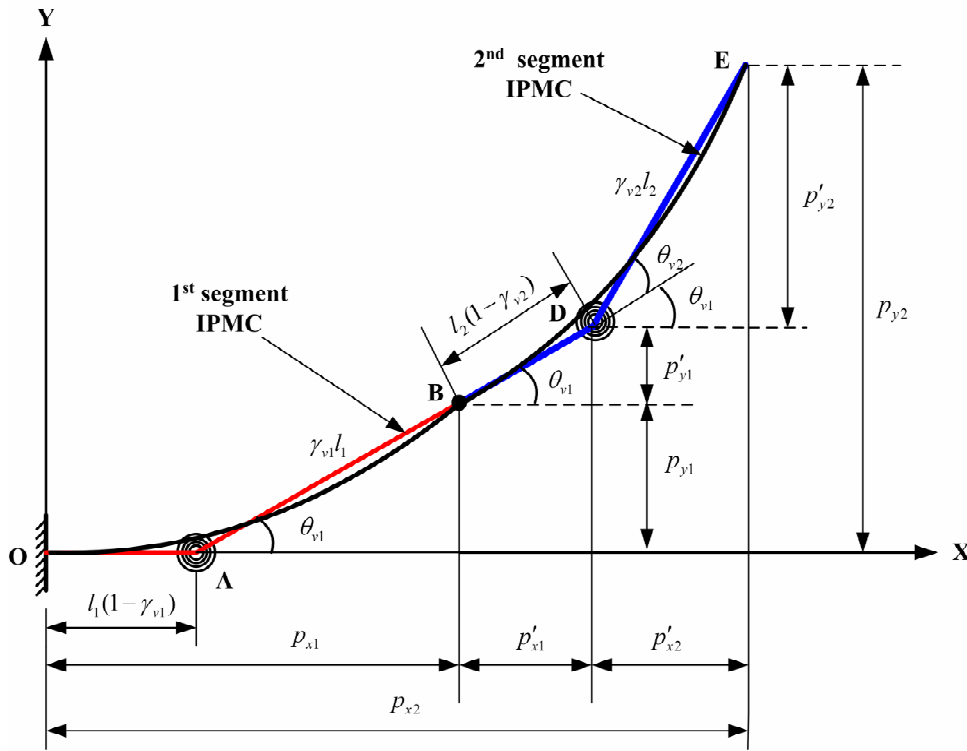
Figure 3.19(a) and (b) show the changes in tip position for a single patch IPMC actuator for various input voltages, while Figure 3.20(a) and (b) demonstrate the effect of dehydration on the tip position using VPPRBM technique. It is observed that accurate tip position can be modeled by VPPRBM technique compared to Euler-Bernoulli and fixed parameter pseudo-rigid body modeling technique.

3.4.2.2 Segmented IPMC Actuator following VPPRBM Technique

Position vector of any point B on the first segment IPMC (p_{x1}, p_{y1}) as shown in

Figure 3.21 is given by:

$$\begin{aligned} p_{x1} &= l_1(1 - \gamma_{v1}) + \gamma_{v1}l_1 \cos \theta_{v1} \\ p_{y1} &= \gamma_{v1}l_1 \sin \theta_{v1} \end{aligned} \quad (3.32)$$



OA=BD: Fixed segment , AB=DE: Rotating segment

Figure 3.21 Bending of two segmented patches of IPMC for the same sequence of input voltage using pseudo-rigid body model.

Similarly, position vector of any point E on the second segment (p_{x2}, p_{y2}) can be expressed as,

$$\begin{aligned} p_{x2} &= p_{x1} + p'_{x1} + p'_{x2} \\ &= l_1(1 - \gamma_{v1}) + \gamma_{v1}l_1 \cos \theta_{v1} + l_2(1 - \gamma_{v2}) \cos \theta_{v1} + \gamma_{v2}l_2 \cos(\theta_{v1} + \theta_{v2}) \end{aligned} \quad (3.33)$$

$$\begin{aligned}
 p_{y2} &= p_{y1} + p'_{y1} + p'_{y2} \\
 &= \gamma_{v1} l_1 \sin \theta_{v1} + l_2 (1 - \gamma_{v2}) \sin \theta_{v1} + \gamma_{v2} l_2 \sin(\theta_{v1} + \theta_{v2})
 \end{aligned}
 \quad (3.34)$$

where, θ_{v1} and $(\theta_{v1} + \theta_{v2})$ are the pseudo-rigid body angles of the first and second segment IPMC respectively with respect to base axis. Assuming identical properties for both the segments, the tip position of the 2nd segment can be obtained as:

$$\begin{aligned}
 p_{x2} &= l_1 - \gamma_{v1} l_1 + l_2 \cos \theta_{v1} + \gamma_{v2} l_2 \cos(\theta_{v1} + \theta_{v2}) \\
 p_{y2} &= l_2 \sin \theta_{v1} + \gamma_{v2} l_2 \sin(\theta_{v1} + \theta_{v2})
 \end{aligned}
 \quad (3.35)$$

Figure 3.22(a) and (b) demonstrating the change in tip position for segmented IPMC actuator using VPPRBM technique.

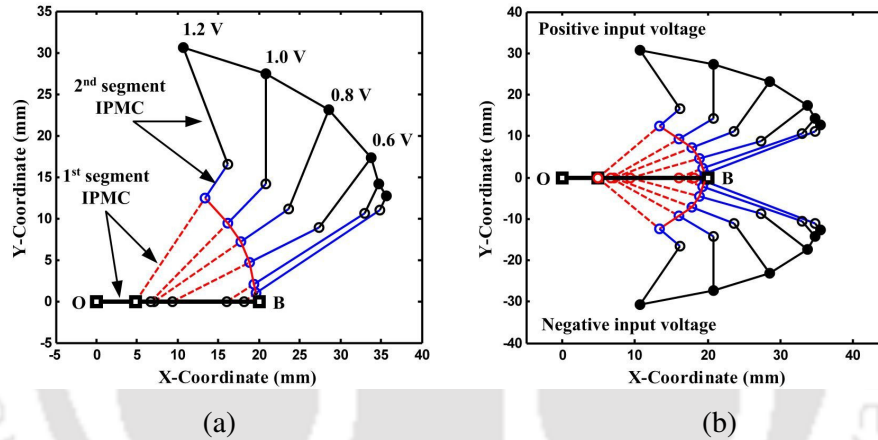


Figure 3.22 Two IPMC patches are bending (a) positive (b) both positive and negative sequence of input voltages following variable parameters pseudo-rigid body model.

3.5 Bending Response Analysis Using VPPRBM Technique

Energy based dynamic model of the segmented patches has been derived using Lagrange principle taking into account loss due to dehydration. Simulation has been performed for the two segmented patches based on the experimental data and the results are shown. The results demonstrate the gradual change in bending resistance, tip position and the bending response due to dehydration.

3.5.1 Dynamic Model of the Segmented IPMC Actuator

Segmented IPMC actuators are modelled similar to serial link robotic manipulator. It is anticipated that, by segmenting IPMC strip in many parts (patches) and applying

different sequence of voltage on different strips, one can control the tip deflection in a way as shown in Figure 3.10. For the segmented IPMC actuator referring to the Figure 3.21, the kinetic energy of the system is the sum of the kinetic energy of the two strips and kinetic energy of the payload. Velocity at point B of the first (v_1) and at point E of the second (v_2) link (Figure 3.21) is obtained by differentiating equations (3.32) and (3.35) i.e.

$$v_1^2 = \dot{p}_{x1}^2 + \dot{p}_{y1}^2 = (\gamma_{v1} l_1)^2 \dot{\theta}_{v1}^2 \quad (3.36)$$

$$v_2^2 = \dot{p}_{x2}^2 + \dot{p}_{y2}^2 = l_2^2 \dot{\theta}_{v1}^2 + (\gamma_{v2} l_2)^2 (\dot{\theta}_{v1} + \dot{\theta}_{v2})^2 \quad (3.37)$$

Neglecting the payload, total kinetic energy of the system is obtained by summing up the kinetic energy of two strips, i.e.

$$T = \frac{1}{2} I_m v_1^2 + \frac{1}{2} I_m v_2^2 \quad (3.38)$$

where, I_m is the area moment of inertia of the link. Substituting the value of (v_1, v_2) into the equation (3.38), the kinetic energy of the system is obtained as:

$$T = \left\{ \frac{1}{2} \rho A l_1 (\gamma_{v1} l_1)^2 + \frac{1}{2} \rho A l_2 (l_2)^2 \right\} \dot{\theta}_{v1}^2 + \left\{ \frac{1}{2} \rho A l_2 (\gamma_{v2} l_2)^2 \right\} (\dot{\theta}_{v1} + \dot{\theta}_{v2})^2 \quad (3.39)$$

Letting, $I_{11} = \rho A l_1 (\gamma_{v1} l_1)^2 + \rho A l_2 (l_2)^2$, $I_{22} = \rho A l_2 (\gamma_{v2} l_2)^2$ and then incorporating this into the equation (3.39),

$$T = \frac{1}{2} I_{11} \dot{\theta}_{v1}^2 + \frac{1}{2} I_{22} (\dot{\theta}_{v1} + \dot{\theta}_{v2})^2 \quad (3.40)$$

where, l_1, l_2 are the lengths of the respective segment.

Potential Energy of the System

The potential energy of the system is due to resistance to deformation of the link, payload and its own weight. Hence, the potential energy of the link can be expressed as $U = U_g + U_s$. Where, U_g and U_s are potential energies due to gravity and torsional spring. Torsional spring of IPMC signifies the bending resistance of IPMC. Neglecting the potential energy due to gravity and payload, the total potential energy of the system can be expressed as:

$$U = U_s = \frac{1}{2} K_{v1} \theta_{v1}^2 + \frac{1}{2} K_{v2} \theta_{v2}^2 \quad (3.41)$$

where, K_{v1}, K_{v2} are the variable torsional springs constant for the 1st and 2nd segment of IPMC patches.

Governing Equation of Motion of the System

The Lagrangian of the system is expressed in terms of kinetic and potential energy i.e.

$$L = T - U$$

$$= \left[\frac{1}{2} I_{11} \dot{\theta}_{v1}^2 + \frac{1}{2} I_{22} (\dot{\theta}_{v1} + \dot{\theta}_{v2})^2 \right] - \left[\frac{1}{2} K_{v1} \theta_{v1}^2 + \frac{1}{2} K_{v2} \theta_{v2}^2 \right] \quad (3.42)$$

Therefore, equation of motion of the system is given by

$$\frac{d}{dt} \left(\frac{\partial L}{\partial \dot{\theta}_{v_i}} \right) - \frac{\partial L}{\partial \theta_{v_i}} = M_{v_i} \quad (3.43)$$

Substituting the equation (3.42) into equation (3.43), the governing equation of the system is obtained as:

$$(I_{11} + I_{22}) \ddot{\theta}_{v1} + I_{22} \ddot{\theta}_{v2} + K_{v1} \theta_{v1} = M_{v01} (1 - \lambda)$$

$$I_{22} \ddot{\theta}_{v1} + I_{22} \ddot{\theta}_{v2} + K_{v2} \theta_{v2} = M_{v02} (1 - \lambda) \quad (3.44)$$

where, M_{v01} and M_{v02} are the bending moment of 1st and 2nd segment of IPMC patches at zero dehydration, respectively. Incorporating the damping into the system, equation (3.44) is modified to:

$$(I_{11} + I_{22}) \ddot{\theta}_{v1} + I_{22} \ddot{\theta}_{v2} + C_{d1} \dot{\theta}_{v1} + K_{v1} \theta_{v1} = M_{v01} (1 - \lambda)$$

$$I_{22} \ddot{\theta}_{v1} + I_{22} \ddot{\theta}_{v2} + C_{d2} \dot{\theta}_{v2} + K_{v2} \theta_{v2} = M_{v02} (1 - \lambda) \quad (3.45)$$

where, C_{d1}, C_{d2} are the respective damping co-efficient of the IPMC patches in dehydrated condition. The generalized governing equation of the system is given by,

$$I_m \ddot{\Theta} + C_d \dot{\Theta} + K_s \Theta = M_{v0} \quad (3.46)$$

$$\text{where, } I_m = \begin{bmatrix} I_{11} + I_{22} & I_{22} \\ I_{22} & I_{22} \end{bmatrix}, C_d = \begin{bmatrix} C_{d1} & 0 \\ 0 & C_{d2} \end{bmatrix}, K_s = \begin{bmatrix} K_{v1} & 0 \\ 0 & K_{v2} \end{bmatrix},$$

$$M_{v0} = \begin{bmatrix} M_{v01}(1-\lambda) \\ M_{v02}(1-\lambda) \end{bmatrix}, \Theta = \begin{bmatrix} \theta_{v1} \\ \theta_{v2} \end{bmatrix}$$

3.6 Results and Discussions

Results are obtained through numerical simulation by solving the differential equation (3.45) in state-space form and are demonstrated. All the results show the variation of tip response at peak bending deflection.

Figure 3.23 shows the change in angular position of the first segment of IPMC actuator for positive and zero dehydration for maximum input 1.2V while Figure 3.24 shows the rate of change in variable pseudo-rigid body angle of the first segment IPMC actuator. Figure 3.25 and 3.26 are showing the rate of change in angular position and variable pseudo-rigid body angle of the second segment IPMC actuator respectively. It is observed that response of the second patch is irregular than first patch to some extent. This may be the effect of coupled motion due to the first segment. The response of the system primarily depends on the variable spring constant of the IPMC actuator. It is further anticipated that as the IPMC actuator gets dehydrated, it becomes stiffer compared to fully hydrated state. It is understood that both the patches response remain uniform throughout the time span for each input voltage.

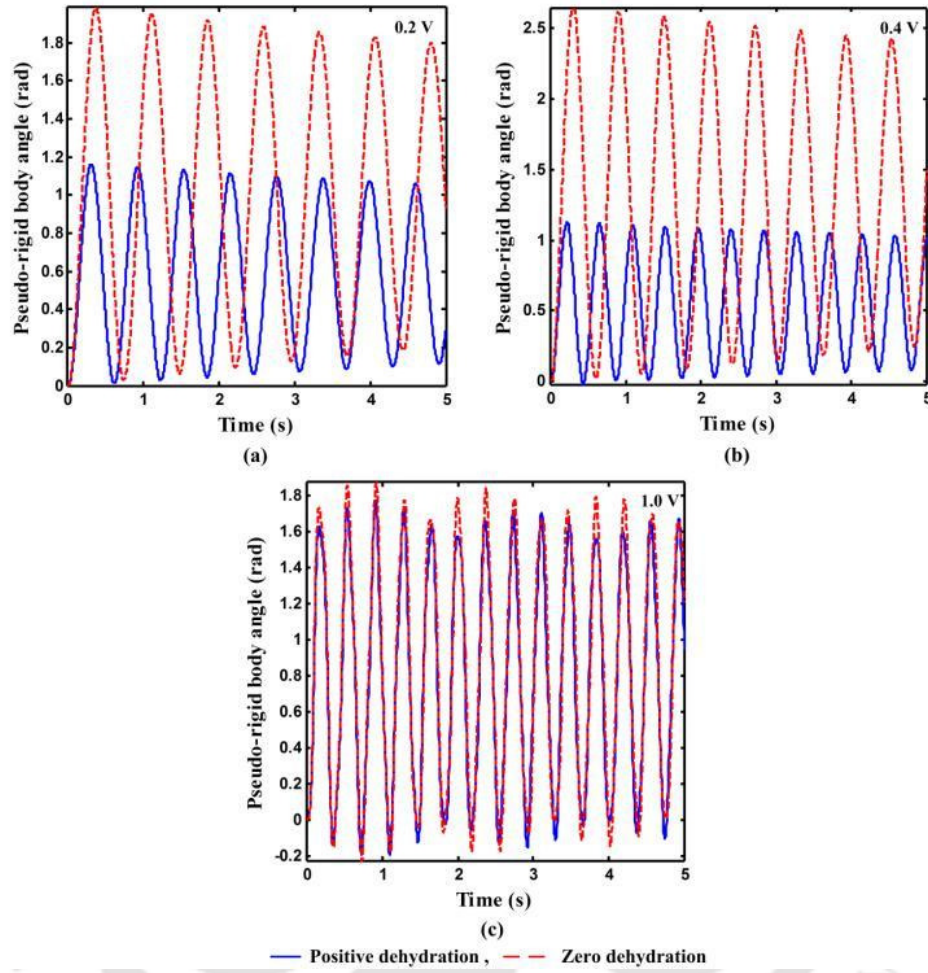


Figure 3.23 Change in angular position of the first segment of IPMC patch with (a) 0.2 V (b) 0.4 V (c) 1.0 V.

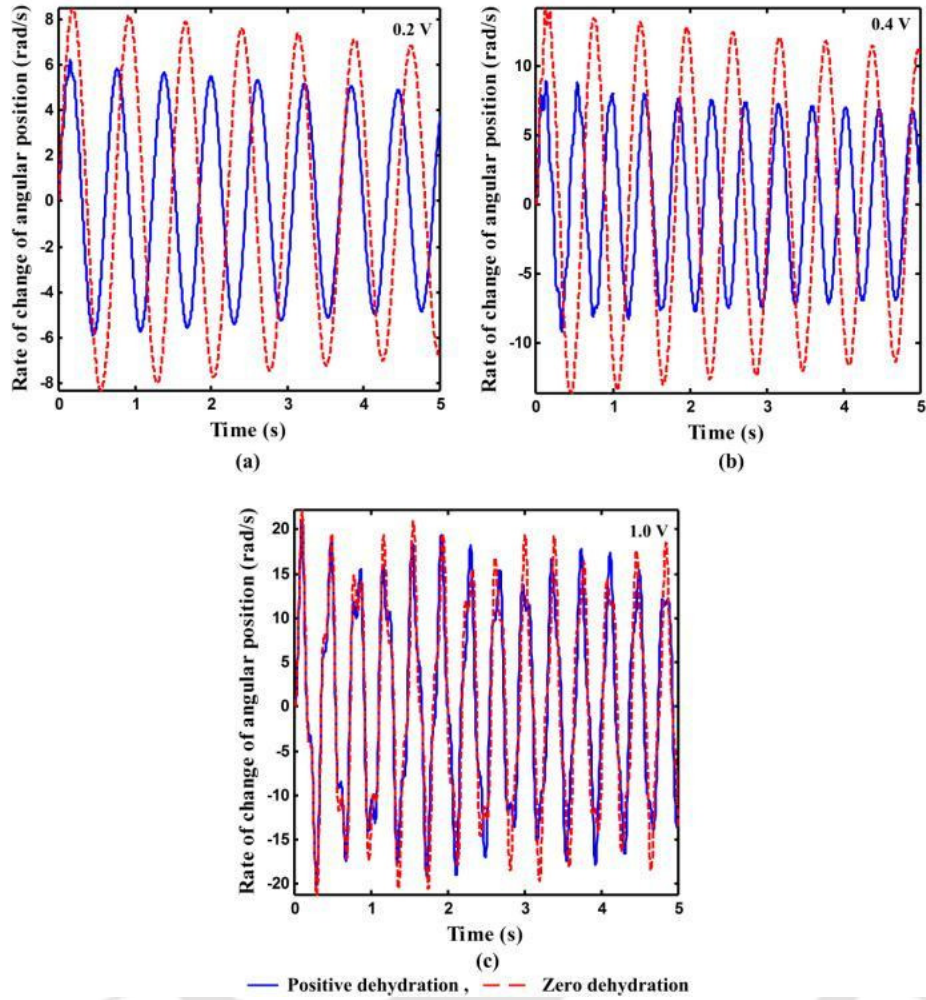


Figure 3.24 Rate of change of angular position of the first segment of IPMC patch with (a) 0.2 V (b) 0.4 V (c) 1.0 V.

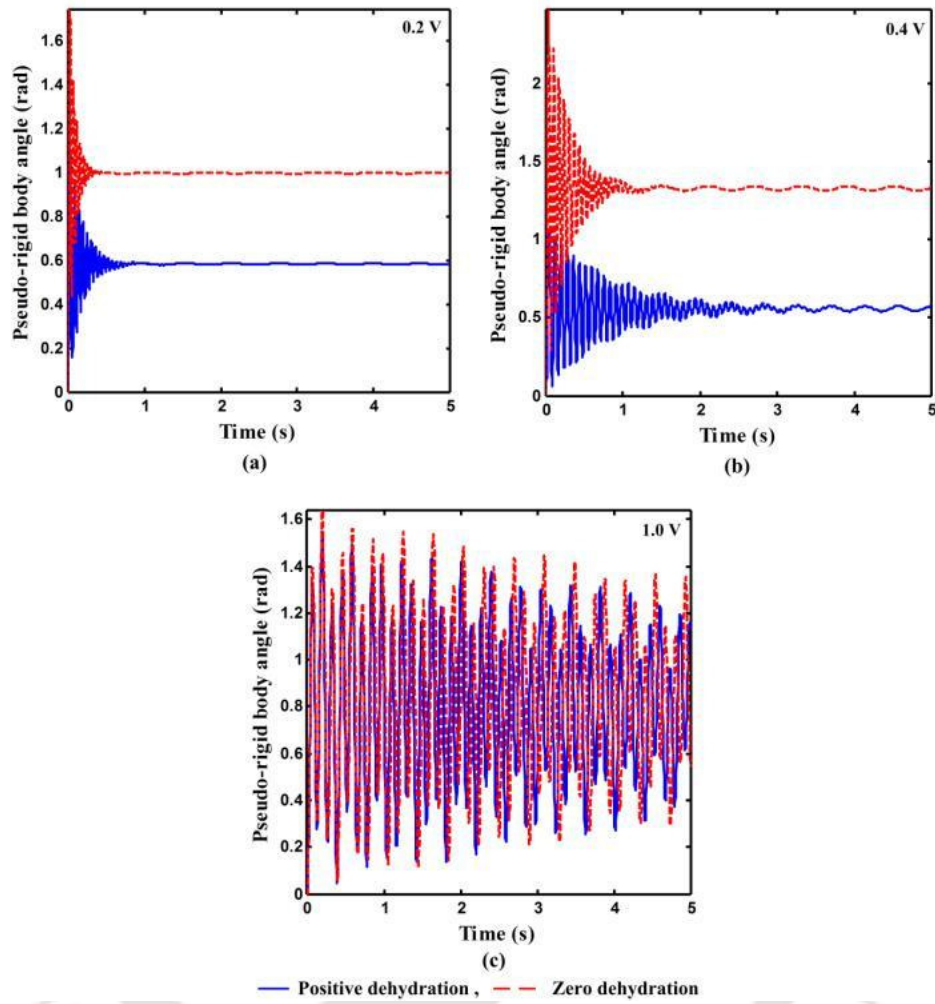


Figure 3.25 Change in angular position of the second segment of IPMC patch with (a) 0.2 V (b) 0.4 V (c) 1.0 V.

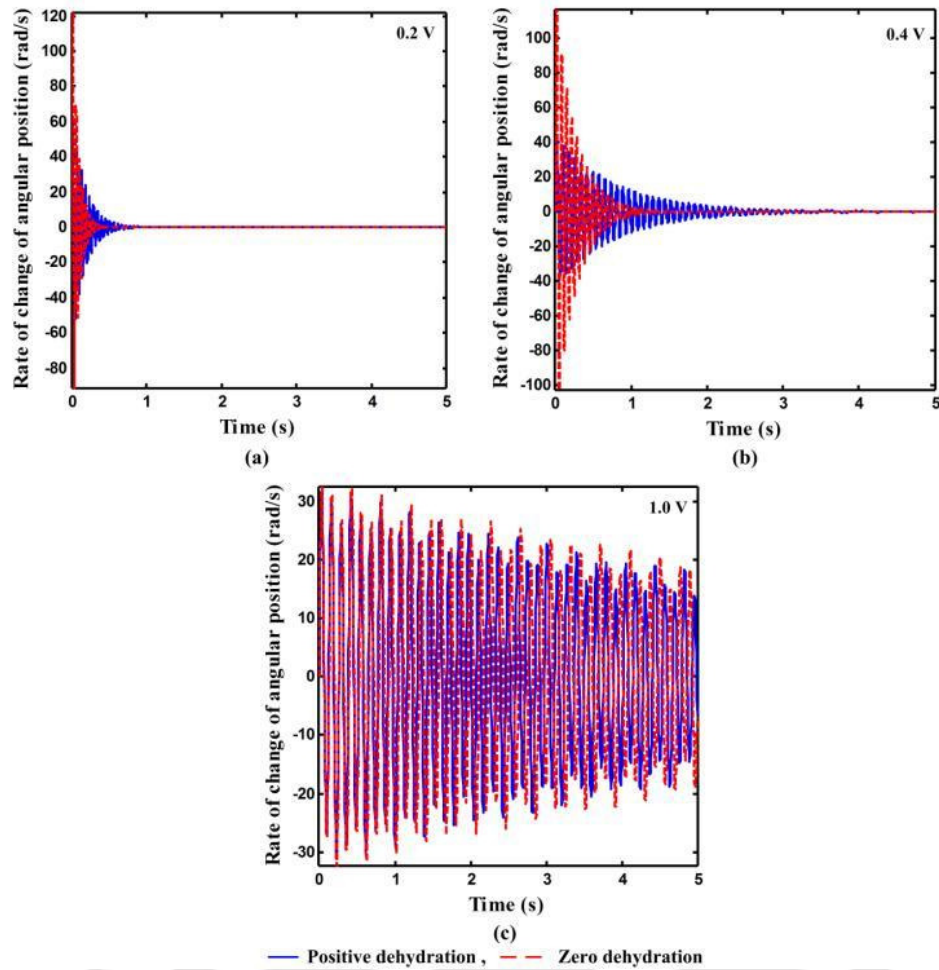


Figure 3.26 Rate of change of angular position of the second segment of IPMC patch with (a) 0.2 V (b) 0.4 V (c) 1.0 V.

3.7 Summary

In this chapter, loss of moisture content of the IPMC actuator is modelled by introducing a new term 'loss-factor' in terms of input voltage and time. As dehydration occurs during the actuation process, the bending tip deflection experimental data are used to establish a power fit correlation for loss-factor in terms of input voltage and time using Cobb-Douglas production method. The proposed model is robust and valid for any IPMC actuator in working medium that experiences continuous dehydration.

In section 3.4, modeling of bending characteristics following Euler-Bernoulli and pseudo-rigid body modeling approach for the single and segmented actuator is formulated after constituting the homogeneous coordinate transformation matrix, assuming it as a serial link multi-degree of freedom manipulator. An energy-based dynamic model of the patches has been derived using the Lagrange principle. Simulations are performed for single and two segmented IPMC patches to demonstrate the bending response for various input voltages. The results demonstrate the gradual reduction of bending response of an actuator owing to moisture loss.

In section 3.5, segmented ionic polymer metal composite actuators are analyzed following the variable parameters pseudo-rigid body modeling (VPPRBM) technique in order to assess the effect of dehydration on bending resistance and bending response. Energy based dynamic model of the patches has been derived after developing the forward kinematics incorporating loss due to dehydration. Simulation has been performed for two segmented IPMC patches to demonstrate the change in bending resistance, bending response and end-tip positioning for various input voltages. It is observed that with dehydration, the bending response of the actuator becomes faster to reach the steady-state although the amplitude of the bending response decreases. The estimation helps to understand the amount of input voltage or the moisture quantity that has to be added with time to compensate the loss. The proposed idea can be employed for controlling the tip position by compensating the loss.

Vibration Characteristics of Silver Electroded IPMC Actuator Subjected to DC Input Voltage

4.1 Introduction

Theoretical and experimental investigation of silver electroded IPMC actuator is carried out to study the transient and steady-state response, stability and accuracy of the system performance subjected to DC potential. Past researches established the fact that an IPMC actuator dehydrates continuously in open environment reduces the initial moisture content with applied electric potential and this eventually affect significantly the actuation performance of IPMC. However, very limited research has been carried out to address the effect of dehydration on the vibration characteristics of IPMC. It is proven fact that during actuation, IPMCs experience low amplitude vibration during bending motion, and hence it is important to study the vibration characteristics. The objectives of the present study is to

- analyze the vibration characteristics of IPMC actuator under DC potential gradient.
- demonstrate theoretically and experimentally the effect of dehydration on vibration response.

Theoretical modelling of the actuator is done using multi-mode approximation; assuming fixed-free Euler-Bernoulli beam and the governing equation of motion of the actuator has been derived by using the extended Hamilton's principle incorporating loss due to dehydration. The obtained multi-degree of freedom system equation has been solved numerically to obtain the transient and steady-state vibration response. Several experiments are conducted to investigate the vibration

characteristics and to validate the theoretical results of the system with 'positive' and 'zero' dehydration.

4.2 Modeling of the IPMC Actuator

Figure 4.1 shows a fixed-free configured IPMC actuator subjected to DC electric potential at the fixed end across its thickness. The actuator is modelled as a continuous distributed parameter system where the deflection is assumed to be consisting of two components, namely displacement due to rigid body motion ($x\theta$) and displacement due to the flexibility of the structure (w) as shown in Figure 4.1. As shown in Figure 4.1, w is the transverse displacement at any location ' x '. Extended Hamilton's Principle is applied for deriving the governing equation of motion of the system and is discussed subsequently in the following subsections.

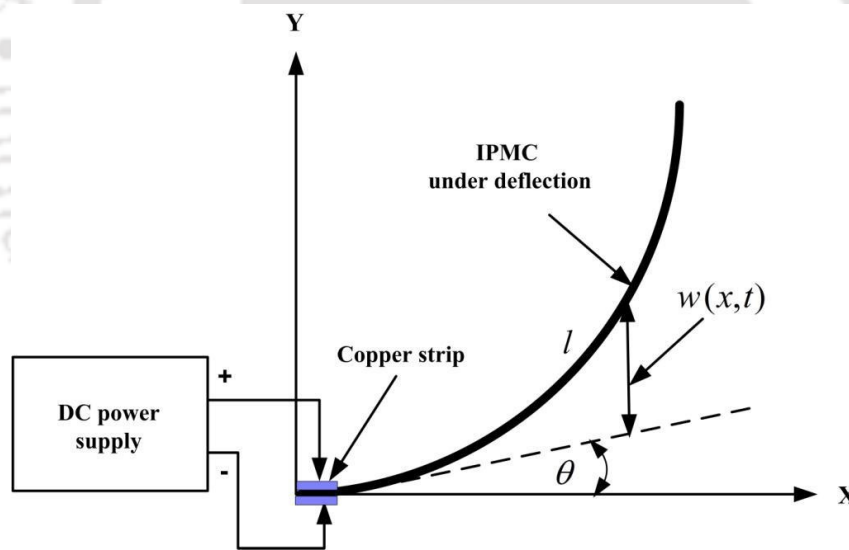


Figure 4.1 An IPMC actuator bending configuration.

4.2.1. Kinetic Energy of the System

Velocity (v) at any point at a distance x along the link can be expressed as:

$$v = x \dot{\theta} + \dot{w} \quad (4.1)$$

where, θ is the angle between the undeformed IPMC actuator (X-axis) and the tangent to the deformed actuator at zero slope as shown in Figure 4.1. It may be noted that a similar modelling approach was also found in the literature by Wang and Vidyasagar (1989); Bandopadhyaya *et al.*, (2007). But in those cases dehydration effects have not been considered and only single-mode approximation is taken in their analysis. The kinetic energy of the actuator (T_a) can be expressed as:

$$T_a = \frac{1}{2} \int_0^l \rho A (x\dot{\theta} + \dot{w})^2 dx = \frac{\rho A}{2} \int_0^l (x^2 \dot{\theta}^2 + 2x\dot{\theta}\dot{w} + \dot{w}^2) dx \quad (4.2)$$

Here, ρ is the mass density of the IPMC material, and A is the area of cross section of the link. Taking multi-mode approximation w can be written as follows.

$$w(x, t) = \sum_{i=1}^n \psi_i(x) q_i(t) \quad (4.3)$$

Here, n is the number of participating modes in the approximation, $\psi_i(x)$ and $q_i(t)$ are the admissible function and time modulation respectively. The admissible function $\psi_i(x)$ is considered to be same as the eigenfunction of a cantilever beam which is given by:

$$\psi_i(x) = -\frac{(\cos(\beta l)_i + \cosh(\beta l)_i)}{(\sin(\beta l)_i + \sinh(\beta l)_i)} (\sin(\beta x)_i - \sinh(\beta x)_i) + (\cos(\beta x)_i - \cosh(\beta x)_i) \quad (4.4)$$

where, $i = 1, 2, 3, \dots, n$ and βl can be obtained from the characteristic equation

$$\cos \beta l \cosh \beta l = -1 \quad (4.5)$$

For the first three modes, $(\beta l)_1 = 1.8751$, $(\beta l)_2 = 4.6941$ and $(\beta l)_3 = 7.8548$.

Substituting the expression of, Equation (4.3) in the Equation (4.2), and arranging the coefficients of θ and q , kinetic energy of the actuator can be written as:

$$T_a = \frac{1}{2} \rho A \frac{l^3}{3} \dot{\theta}^2 + \rho A \int_0^l x \sum_{i=1}^n \psi_i(x) \dot{q}_i \dot{\theta} dx + \frac{1}{2} \rho A \int_0^l x^2 \sum_{i=1}^n \sum_{j=1}^n \dot{q}_i \psi_i \dot{q}_j \psi_j dx \quad (4.6)$$

$$\text{or, } T_a = \frac{1}{2} I_{\theta\theta} \dot{\theta}^2 + \sum_{i=1}^n I_{iq_{ii}} \dot{\theta} \dot{q}_i + \frac{1}{2} \sum_{i=1}^n \sum_{j=1}^n \dot{q}_i I_{qq_{ij}} \dot{q}_j \quad (4.7)$$

where, $I_{\theta\theta} = \rho A \frac{l^3}{3}$, $I_{iq_{ii}} = \rho A \int_0^l x \psi_i(x) dx$, $I_{qq_{ij}} = \int_0^l \psi_i(x) \rho A \psi_j(x) dx$

Figure 4.2 shows the first three mode shapes of vibration of the actuator.

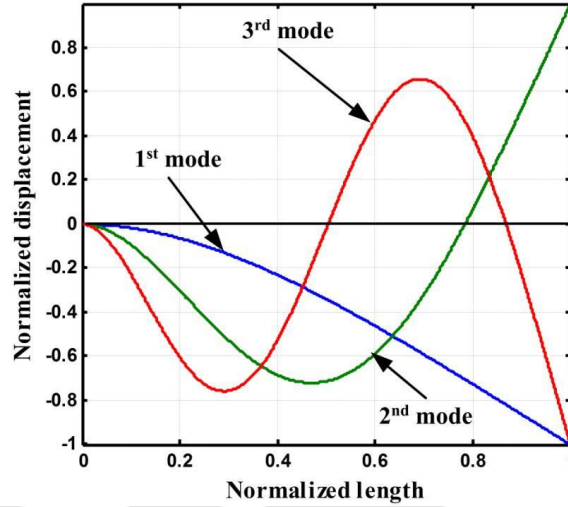


Figure 4.2 First three mode shapes of vibration.

4.2.2. Potential Energy of the System

Neglecting gravitational effect, the potential energy of the system is considered due to the deformation of the actuator only and is expressed as:

$$V_a = \frac{1}{2} \int_0^l EI \left[\frac{\partial^2 w}{\partial x^2} \right]^2 dx \quad (4.8)$$

Substituting equation (4.3) into (4.8) one may write,

$$\begin{aligned} V_a &= \frac{1}{2} \int_0^l EI \left[\frac{\partial^2 w}{\partial x^2} \right]^2 dx = \frac{1}{2} \int_0^l EI \left[\sum_{i=1}^n \frac{\partial^2 \psi_i(x)}{\partial x^2} q_i(t) \right]^2 dx \\ &= \sum_{i=1}^n \sum_{j=1}^n q_i^T \left[\frac{1}{2} \int_0^l EI \frac{\partial^2 \psi_i}{\partial x^2} \frac{\partial^2 \psi_j}{\partial x^2} dx \right] q_j = \frac{1}{2} \sum_{i=1}^n \sum_{j=1}^n q_i^T K_{ij} q_j \end{aligned} \quad (4.9)$$

$$\text{where, } K_{ij} = \int_0^l \frac{\partial^2 \psi_i}{\partial x^2} EI \frac{\partial^2 \psi_j}{\partial x^2} dx$$

4.2.3. Work done due to Damping and Input Voltage

Work done due to damping of the actuator can be expressed as:

$$W_c = \frac{1}{2} \int_0^l C \dot{w}^2 dx. \quad (4.10)$$

Here, C is the equivalent viscous damping coefficient assumed in this work. Substituting equation (4.3) into (4.10) one may write,

$$W_c = \frac{1}{2} C \int_0^l \left[\sum_{i=1}^n \psi_i(x) \dot{q}_i(t) \right]^2 dx = \frac{1}{2} \sum_{i=1}^n \sum_{j=1}^n \dot{q}_i^T C_{ij} \dot{q}_j. \quad (4.11)$$

$$\text{where, } C_{ij} = \int_0^l \psi_i(x) C \psi_j(x) dx$$

Further, work done by IPMC actuator is due to the bending moment generated by the applied electric potential. Assuming an equivalent force $F(x, V, t)$ acting at the tip of the actuator (at $x = l$) which causes the same bending moment M due to the applied voltage, the work done due to infinitesimal bending $\delta\theta$ can be written as follows.

$$\delta W_a = M \delta\theta = M_0(1 - \lambda) \delta\theta = (F \times p_x) \delta\theta \quad (4.12)$$

where, λ is the loss-factor, the expression of which is derived in chapter 3. Also, M, M_0 are the bending moments for ‘positive’ and ‘zero’ dehydration respectively. The force $F(x, V, t)$ depends on the applied voltage and the material properties of the IPMC actuator as discussed in chapter 3. For fully hydrated system $\lambda = 0$.

4.2.4. Equation of Motion of the System

To derive the governing equation of motion, the expressions for total kinetic energy T_a , potential energy V_a and work done due to damping δW_c and due to applied voltage δW_a are used in the following equation for extended Hamilton’s principle

$$\int_{t_1}^{t_2} (\delta T_a - \delta V_a - \delta W_c + \delta W_a) dt = 0 \quad (4.13)$$

to yield the following equation.

$$\begin{aligned} & \int_{t_1}^{t_2} \left[-I_{tt} \ddot{\theta} - \sum_{i=1}^n I_{tq_{ii}} \ddot{q}_i + M_0(1-\lambda) \right] \delta\theta dt \\ & + \int_{t_1}^{t_2} \left[-I_{tq_{ii}} \ddot{\theta} - \sum_{j=1}^n I_{qq_{ij}} \ddot{q}_j - \sum_{j=1}^n C_{ij} \dot{q}_j - \sum_{j=1}^n K_{ij} q_j \right] \delta q_i dt = 0 \end{aligned} \quad (4.14)$$

As $\delta\theta$ and δq_i ($i=1, 2, \dots, n$) are the virtual displacements and can take any value, the terms in the square brackets are set to zero to get the equation of motions of the system i.e.,

$$\begin{aligned} I_{tt} \ddot{\theta} + \sum_{i=1}^n I_{tq_{ii}} \ddot{q}_i &= M_0(1-\lambda) \\ I_{tq_{ii}} \ddot{\theta} + \sum_{j=1}^n I_{qq_{ij}} \ddot{q}_j + \sum_{j=1}^n C_{ij} \dot{q}_j + \sum_{j=1}^n K_{ij} q_j &= 0, i=1, 2, \dots, n \end{aligned} \quad (4.15)$$

These $n+1$ equations of motion can be written in terms of mass (I_s), damping (C_s) and stiffness (K_s) matrices as follows.

$$I_s \ddot{\Theta} + C_s \dot{\Theta} + K_s \Theta = M_s \quad (4.16)$$

Taking $n=3$ i.e., three mode approximation, the coefficients of equation (4.16) can be written as follows.

$$\begin{aligned} \text{where, } I_s &= \begin{pmatrix} I_{tt} & I_{tq_{11}} & I_{tq_{22}} & I_{tq_{33}} \\ I_{tq_{11}} & I_{qq_{11}} & 0 & 0 \\ I_{tq_{22}} & 0 & I_{qq_{22}} & 0 \\ I_{tq_{33}} & 0 & 0 & I_{qq_{33}} \end{pmatrix}, C_s = \begin{pmatrix} C_{11} & C_{12} & C_{13} & C_{14} \\ C_{21} & C_{22} & C_{23} & C_{24} \\ C_{31} & C_{32} & C_{33} & C_{34} \\ C_{41} & C_{42} & C_{43} & C_{44} \end{pmatrix}, \\ K_s &= \begin{pmatrix} 0 & 0 & 0 & 0 \\ 0 & K_{11} & 0 & 0 \\ 0 & 0 & K_{22} & 0 \\ 0 & 0 & 0 & K_{33} \end{pmatrix}, M_s = \begin{pmatrix} M_0(1-\lambda) \\ 0 \\ 0 \\ 0 \end{pmatrix}, \Theta = \begin{pmatrix} \theta \\ q_1 \\ q_2 \\ q_3 \end{pmatrix} \end{aligned}$$

Equation (4.16) is solved using RK4 method to obtain θ, q_1, q_2, q_3 which can be further used to obtain the position of the actuator at a given voltage. Damping matrix is considered to be Rayleigh proportional type and can be expressed as $C_s = \alpha_a I_s + \beta_a K_s$, where α_a and β_a are the parameters obtained experimentally.

4.3 Results and Discussions

4.3.1 Numerical Results

Numerical simulation are performed for a single patch IPMC actuator taking into account of the experimental data as given in Table 3.1 and the physical properties of the IPMC as given in Table 3.4. Unlike in most of the analyses where single mode study is done, in the present work, first three modes have been taken into consideration for analysis. The governing equation of motion of the system (4.16) is solved subsequently to obtain the vibration response for various input voltage. For numerical calculation, α_a and β_a values are taken as 0.0860 and 0.0032, respectively.

All the results are showing the vibration response ‘positive dehydration’ and also for ‘zero dehydration’ condition. In the experiment, tip deflection data are measured in steady-state for various input voltages. Thus in numerical study vibration response data are plotted and the steady-state results are taken for both the cases (positive and zero dehydration) and the results are compared subsequently. The vibration response for various input voltages has been plotted considering first three modes as shown in Figure 4.3. It is observed that for a particular input voltage the high transient response gradually diminishes with time and reaches steady-state. Further, it is also observed that as the input voltage increases, the amplitude of vibration increases rapidly but the response diminishes at a faster rate with time. Figure 4.4 demonstrates the effect of dehydration on the vibration response of the actuator for various input voltages. It is observed that with ‘positive dehydration’ the amplitude of vibration decreases in comparison to hydrated condition.

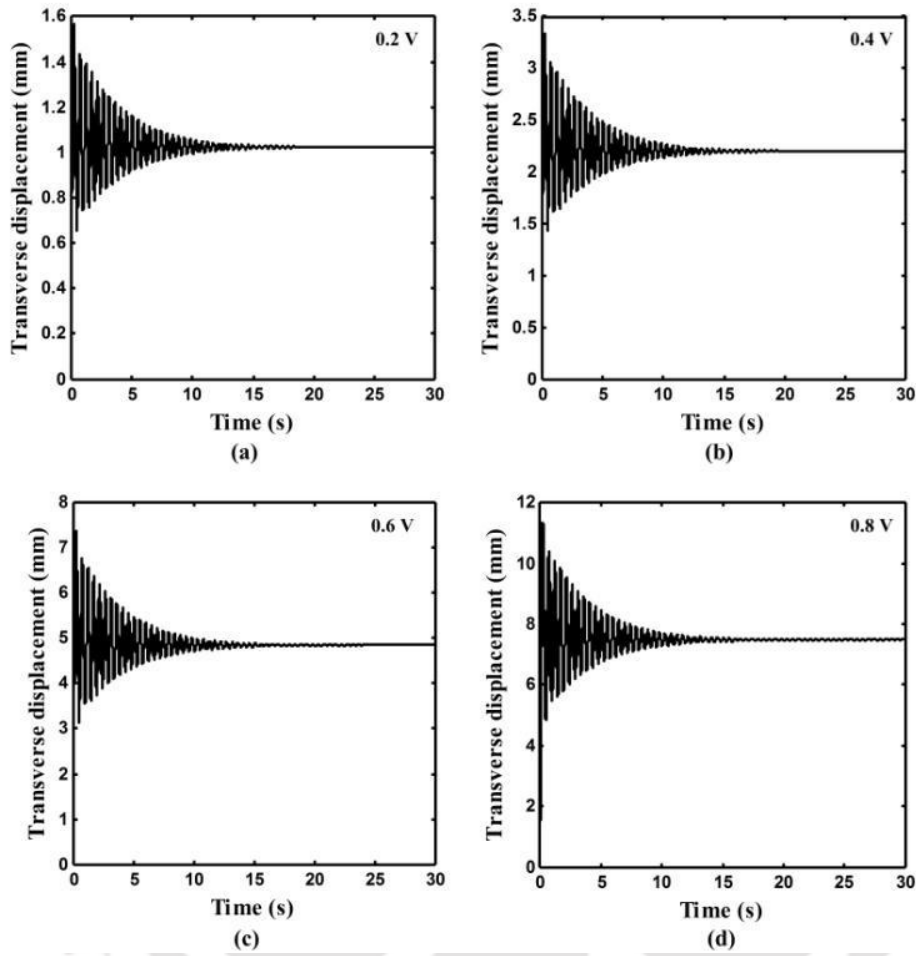


Figure 4.3 Vibration response of the IPMC actuator with various input voltages (a) 0.2V (b) 0.4V (c) 0.6V (d) 0.8V.

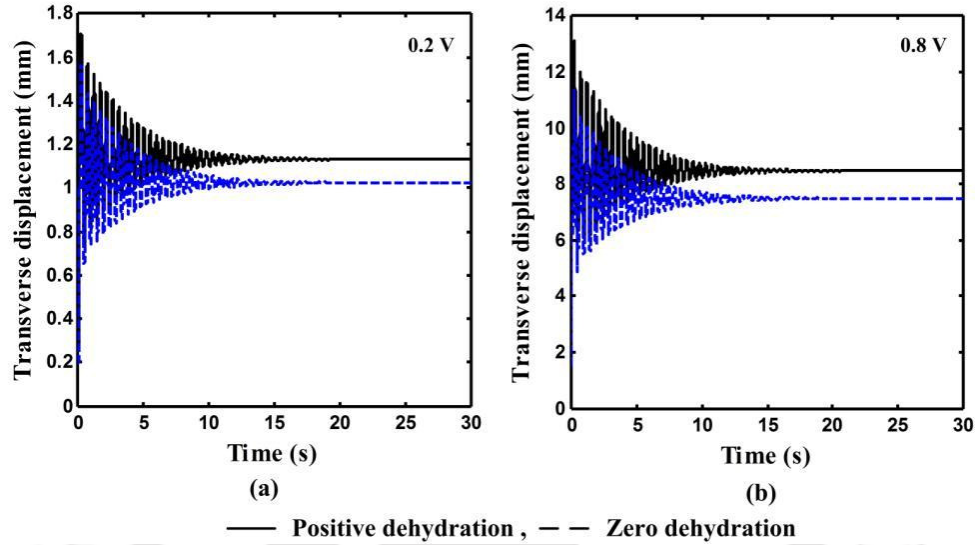


Figure 4.4 Effect of dehydration on the vibration response of the IPMC actuator with various input voltages (a) 0.2V (b) 0.8V.

Figure 4.5 (a) and (b) show comparative study on tip deflection obtained through experimentally and numerically solving equation (4.16). It is observed that (Table 4.1) the percentage error between experimental and simulated results for ‘positive dehydration’ and ‘zero dehydration’ condition at 0.2V is around 2.4% and 2.9%, respectively, while at 0.8V, it is observed to be around 4.0% and 6.0%. Therefore, it is clearly observed that with increase in input voltage, the percentage error of tip deflection increases. This results clearly validate the dehydration model (Equation (3.1)) i.e., with increase in input voltage loss-factor increases. Figure 4.6 shows the percentage change in tip deflection during steady-state for input voltage of 0.2V and 0.8V. It is observed that during steady-state the percentage reduction of amplitude is around 9.5% and 11.6 % for an input of 0.2V and 0.8V, respectively. The results also clearly validate the dehydration model (Equation (3.1)) i.e., loss- factor increases with the input voltage. Hence, the developed model is suitable for finding the response of the actuator easily without going for expensive experiments.

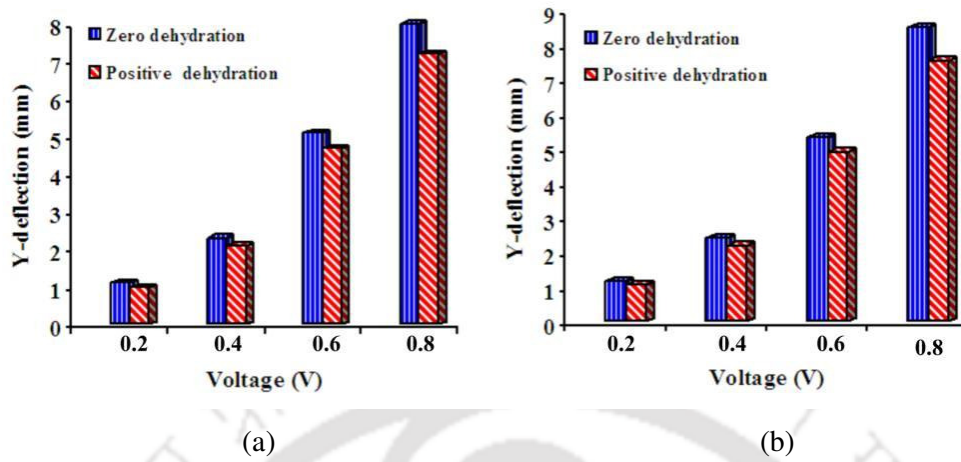


Figure 4.5 (a) Experimental (b) Estimated results for Y-deflection at steady-state.

Table 4.1 Comparison of experimental and simulated results for tip position.

Voltage	Tip position (p_y) at steady-state (mm)					
	Positive dehydration			Zero dehydration		
	Experimental	Theoretical	% error	Experimental	Theoretical	% error
0.2V	1	1.024	2.4	1.1	1.132	2.9
0.4V	2.1	2.166	3.14	2.3	2.386	3.7
0.6V	4.7	4.852	3.23	5.1	5.307	4.05
0.8V	7.2	7.491	4.04	8.0	8.479	6.0

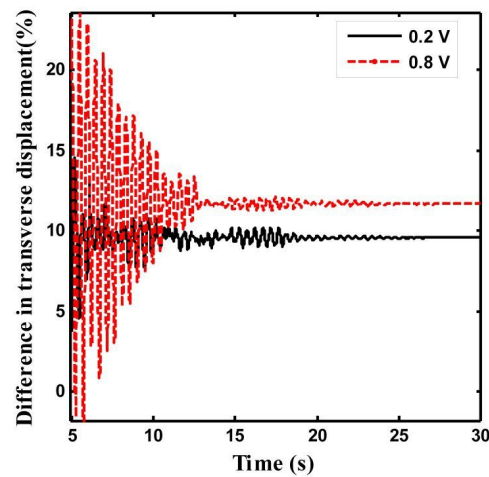


Figure 4.6 Percentage differences in the transverse displacement of IPMC for input voltages of 0.2V and 0.8V for hydrated and dehydrated conditions.

It is also observed that as the voltage increases the reduction increases gradually. Further, it is anticipated that as the IPMC actuator dehydrates continuously in working medium, the transport of mobile cations along with water molecules with them is reduced gradually. Thus the numerical study of the vibration response of the IPMC actuator clearly demonstrates the effect of dehydration on steady-state response and also on the transient response. Both the cases, it is observed that due to dehydration steady-state tip position and also the amplitude of vibration (transient) response decreases.

4.3.2 Experimental Results

An IPMC actuator of size 20x5x0.2 (mm³) is prepared and tested at different time interval under input voltage of 0.2V and 0.4V and the transient vibration characteristics with dehydration are studied. The experimental setup is shown in Figure 4.7. One end of the control module is connected to the Rotational Laser Vibrometer (RLV-5500; make: Polytec, GmbH) and other end to the computer where the data are processed using Bruel and Kjaer PULSE Lab Shop, Version 13.1.0.246 software. The DC power supply (Aplab Limited, 0-32 V DC, 0-2 A) is used to apply the voltage.

Before the experiment, IPMC sample is boiled in deionized water for 10 min and kept within deionized water at room temperature for 24h to ensure full hydration. The entrapped surface moistures are removed by keeping the actuator in open environment for 5min prior to the experiment. Experiments are conducted by applying voltage at the fixed end of the cantilever actuator for 30s and the response of the free end is measured by Rotational Laser Vibrometer. The sample is kept in open environment further for 10 min, 15 min, 30 min and 60 min, successively prior to the next experiment to ensure adequate dehydration. All the experiments are conducted at room temperature. The experimental results are given in Figures 4.8-4.16.

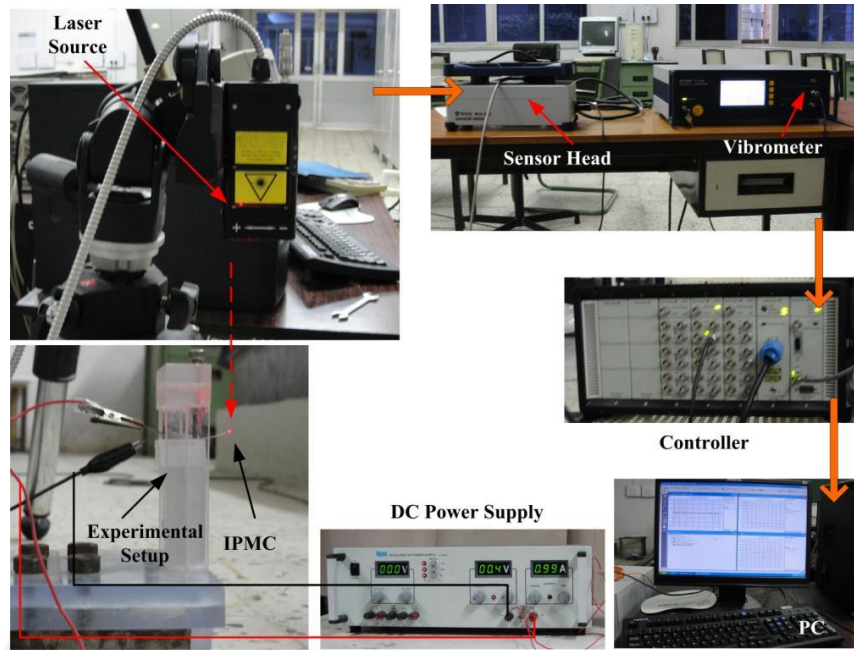


Figure 4.7 Photograph of the experimental setup for vibration analysis.

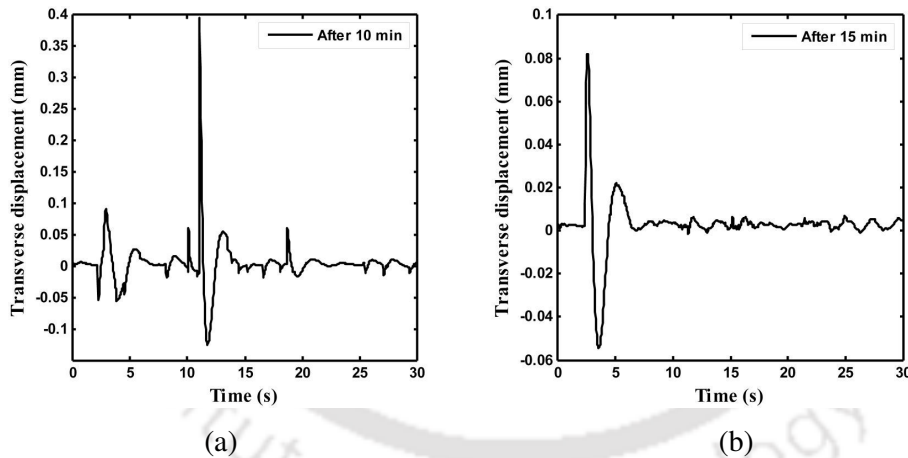


Figure 4.8 Vibration response (a) after 10 min (b) after 15 min for an input of 0.2V.

Figure 4.8(a) and (b) show the vibration response of the actuator after 10 min and 15 min from the initial condition, and the combined response of the actuator up to 30s is shown in Figure 4.9. Figures 4.10 (a) and (b) show the vibration response of the actuator after 30 min and 60 min from the starting of the experiment and it is observed that maximum amplitude reduces further to 50%. It is also observed that due to dehydration transverse displacement of the actuator is reduced considerably. The combined results up to 30s are shown in Figure 4.11. Figure 4.12(a) and (b)

show the corresponding FFT of the vibration response as shown in Figure 4.8(a) and (b), respectively. It is observed that bandwidth of the vibration remains within 2-30 Hz.

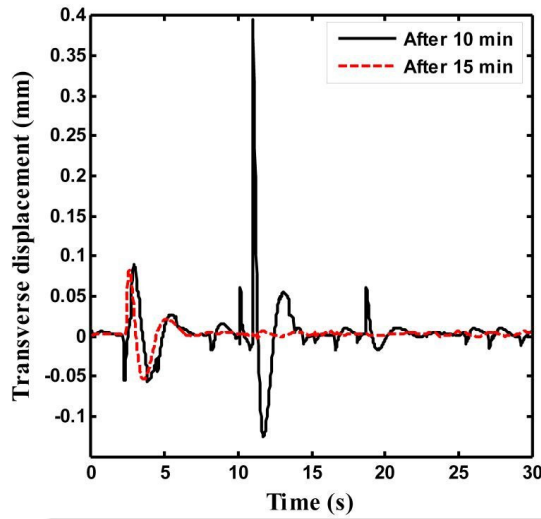


Figure 4.9 Vibration response after 10 min and 15 min for an input of 0.2V.

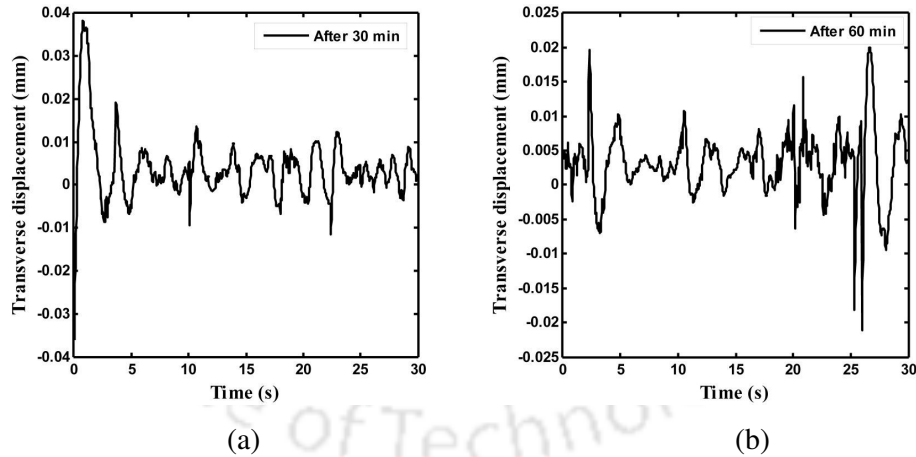


Figure 4.10 Vibration response (a) after 30 min (b) after 60 min for an input of 0.2V.

The experimental results, (Figures 4.9-4.13) demonstrate that the system reaches steady-state after approximately 15-20s. It is due to the fact that as the time interval increases, the loss of moisture content increases and after certain period of time it reaches a saturation limit (IPMC actuator gets dried). The observation validates the loss- factor model, presented in the chapter 3 (Table 3.1 and Figure 3.3). Figure 4.13

illustrates all the response results for a time interval of 10s and compared with the initial (fully hydrated) condition. Thus, it can be concluded that, for a particular input voltage the amplitude of vibration (transverse displacement) of the IPMC actuator gradually decreases due to dehydration effect. Thus the experimental results demonstrate that as the IPMC actuator gets dehydrated, both the transient response (transverse displacement) and the steady-state response (Figure 4.5 (a)) decreases compared to the fully hydrated condition.

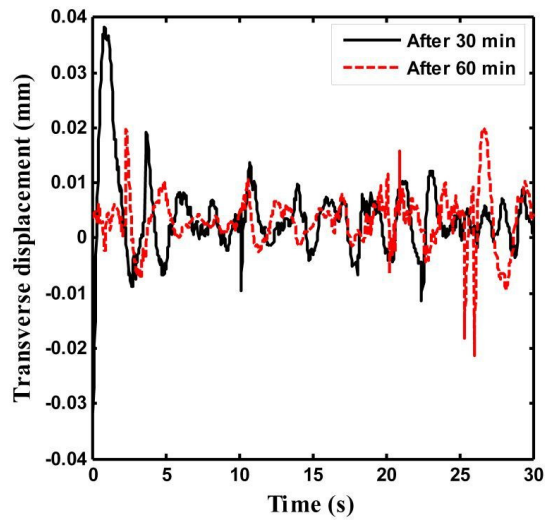


Figure 4.11 Vibration response after 30 min and 60 min for an input of 0.2V.

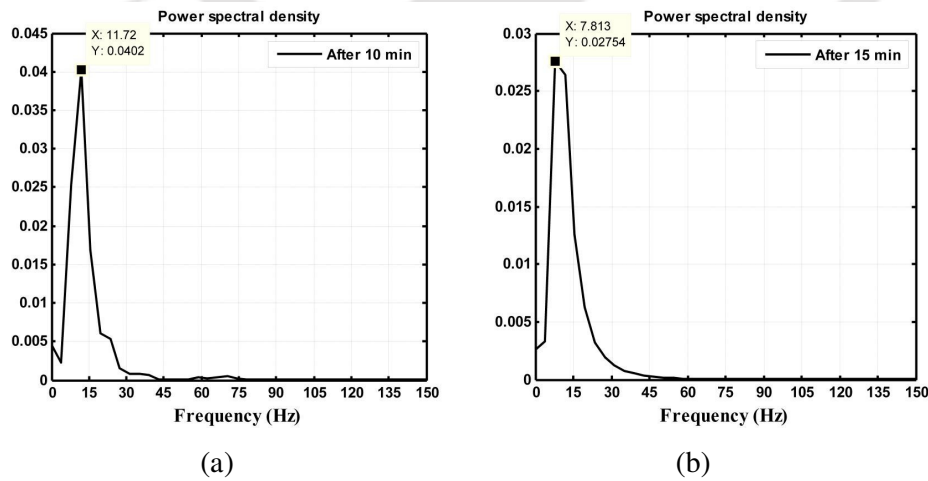


Figure 4.12 FFT (a) after 10 min (b) after 15 min for an input of 0.2V.

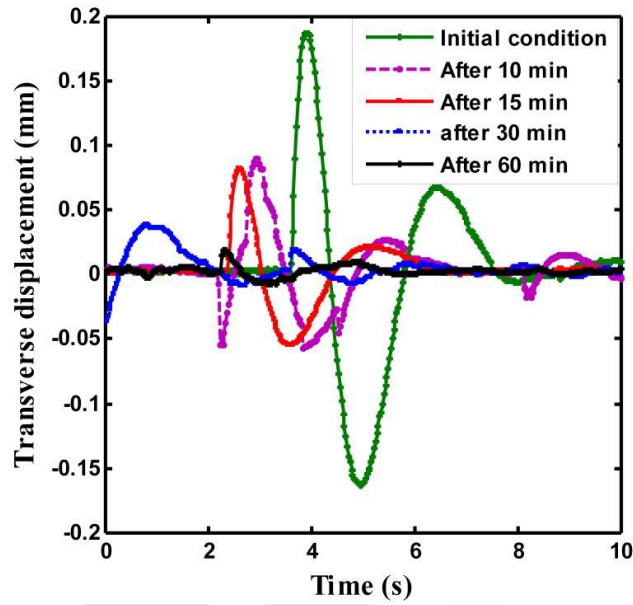


Figure 4.13 Vibration response at different time interval compared to initial condition for an input of 0.2V.

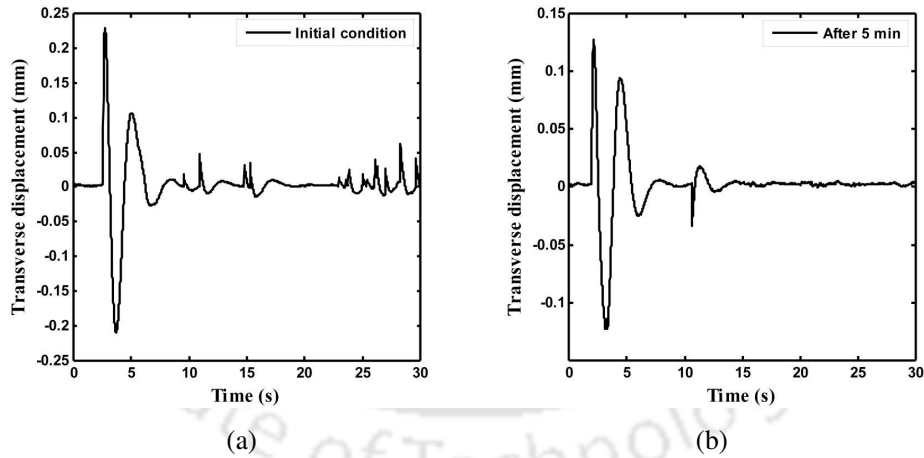


Figure 4.14 Vibration response (a) Initial condition (b) after 5 min for step input of 0.4 V.

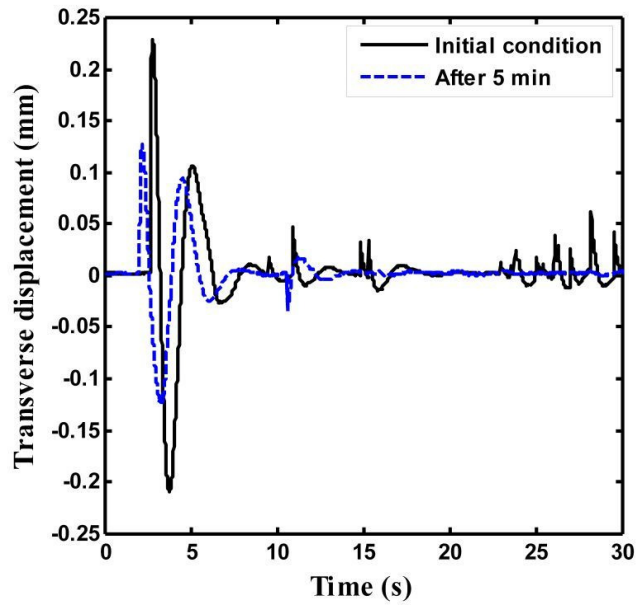


Figure 4.15 Vibration response initial conditions and after 5 min for step input of 0.4 V.

Figures 4.14-4.16 show the vibration response for an input voltage of 0.4V. It is observed that the amplitude of vibration response increases with increase in input voltage. It is also observed that, with 'positive dehydration' the amplitude of vibration decreases compared to hydrated condition.

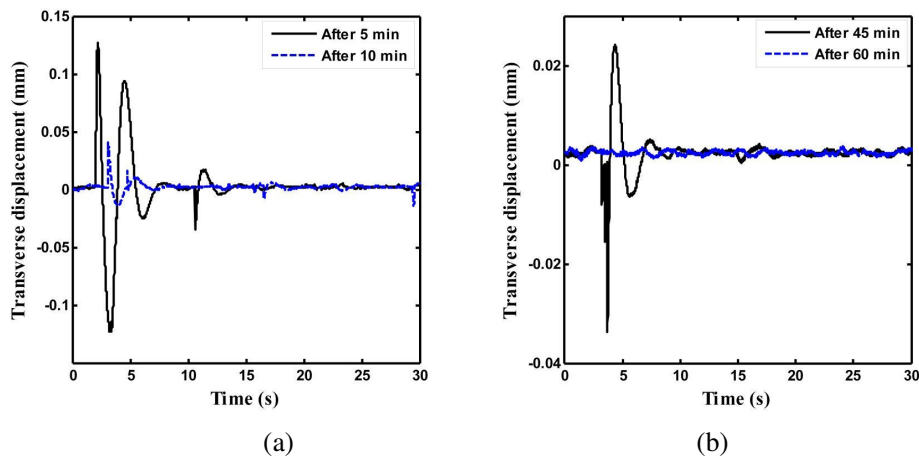


Figure 4.16 Comparison between vibration response (a) after 5 and 10 min (b) after 45 and 60 min for step input of 0.4 V.

4.4. Summary

In this chapter, theoretical modelling of an IPMC actuator is done using multi-mode approximation for studying the vibration response under DC input voltage. Extended Hamilton's principle has been applied to derive the governing equation of motion incorporating loss due to dehydration (loss-factor). The obtained multi-degree of freedom system equation has been solved numerically to obtain the transient and steady-state vibration response. Several experiments are conducted to investigate the vibration characteristics and the results are validated and compared with the loss-factor model. Both theoretical and experimental investigations confirm that both bending tip deflection and amplitude of vibration response decreases with dehydration. It is also observed that the amplitude of vibration response increases with the increase in input voltage. The steady-state response is found to be in good agreement with the experimental results. The proposed dehydration model and experimental results useful for controlling the end-tip position accurately by compensating the loss due to dehydration.

Nonlinear Vibration Analysis of Silver Electroded IPMC Actuator Subjected to Alternating Voltage

5.1 Introduction

The nonlinear characteristics of IPMC actuator is observed to be due to the structural flexibility of the system; which also changes with change in the applied potential difference, polymer property (type of polymer, counter ion, hydration level/moisture/water content), electrode layer (type, area, thickness), and as well as other environmental conditions. Past literatures show that little research has been carried out to find out the nonlinear vibration characteristics of IPMC actuator, excited by a small alternating electric potential. Though development of a comprehensive model is still going on, nonlinear dynamic responses and stability analysis of IPMC have received less attention in the previous work. It is anticipated that an IPMC actuator experiences mechanical vibration during the working period. When the actuator is more flexible and working under high input frequency, a linear dynamic model will no longer be valid. Thus, it is relevant to address and investigate the nonlinear response of IPMC actuator. In this chapter, nonlinear vibration characteristic of an IPMC actuator is analyzed in a fixed-free configuration under alternating electric potential. As the moisture content play an important role in actuation, the system is studied under fully hydrated and dehydrated environmental conditions. Initially, IPMCs are fabricated following the chemical decomposition method using Nafion-117 as the base polymer and non-precious metal silver as the surface electrode. Several experiments are conducted for the IPMC actuator to evaluate its bending and vibration characteristics under alternating potential.

In both hydrated and dehydrated conditions, the actuator is modelled as a flexible distributed parameter system and satisfies Euler-Bernoulli beam

assumptions. The governing equation of motion has been developed using d'Alembert's principle, and generalized Galerkin's method has been used to derive the temporal equation of motion. Method of multiple-scales has been used to solve the temporal equation of motion and subsequently the steady-state response and the stability of the system have been studied. The influence of various system parameters such as amplitude, and frequency of the applied electric potential on the frequency response curves has also been investigated at resonance condition. Simulations have been performed based on experimental data and by solving the temporal equation of motion of the system. Time and frequency response along with phase portrait have been plotted to study the system response. The results obtained by using the perturbation method are compared with the results obtained by numerically solving the temporal equation of motion. Experiment is conducted for a single patch IPMC actuator to evaluate and correlate the vibration response with the theoretical results.

5.2 Nonlinear Mathematical Model of IPMC Actuator

A nonlinear model is developed for an IPMC actuator taking into account of the bending characteristics of the IPMC actuator. This model will be suitable for any IPMC actuator to evaluate and show the nonlinear response. The analysis will help to study the stability of the system under alternating electric potential.

5.2.1 Calculation of Tip Force

To obtain the bending characteristics of an IPMC actuator experiments are conducted by taking IPMC actuator of size $20 \times 5 \times 0.2$ (mm³) in cantilever configuration. Varying input voltage is applied at the fixed end and the bending characteristic is measured subsequently as described section 2.2.4.1. The experimental setup is shown in Figure 2.12. Assuming a constant curvature bending throughout the IPMC sample i.e, $R_c = \frac{l}{\theta}$, where R_c is the radius of curvature, l is

the length of the IPMC, θ is the tip angle and using curve fitting technique the relationship between radius of curvature and applied voltage is obtained as:

For ‘positive’ dehydration, Figure 5.1(a)

$$R_{cd} = 106.3 V^4 - 690.3 V^3 + 1408 V^2 - 1189 V + 384.1, R^2 = 0.9998 \quad (5.1)$$

For ‘zero’ dehydration, Figure 5.1(b)

$$R_{c0} = 129.05 V^4 - 712.52 V^3 + 1353.5 V^2 - 1105 V + 352.81, R^2 = 0.9997 \quad (5.2)$$

Figure 5.1 (a) and (b) show the variation of radius of curvature with applied voltage (V) of the IPMC actuator in dehydrated (positive dehydration) and hydrated (zero dehydration) condition, respectively. R_{cd} represents the radius of curvature with positive dehydration while R_{c0} represents the radius of curvature with zero dehydration. Assuming an equivalent force F acting at the tip of the actuator, the maximum bending moment generated by the actuator can be expressed as $M = F \times p_x$, where p_x is the X-coordinate of tip position of the actuator. Taking into account of pure bending $\frac{M}{I} = \frac{E}{R_c}$ [Popov, (1990)], the tip force (F) developed due to input voltage (V) with positive dehydration can be expressed as:

$$F_d = \frac{EI}{R_{cd} p_x} \quad (5.3a)$$

For zero dehydration condition,

$$F_0 = \frac{EI}{R_{c0} p_x} \quad (5.3b)$$

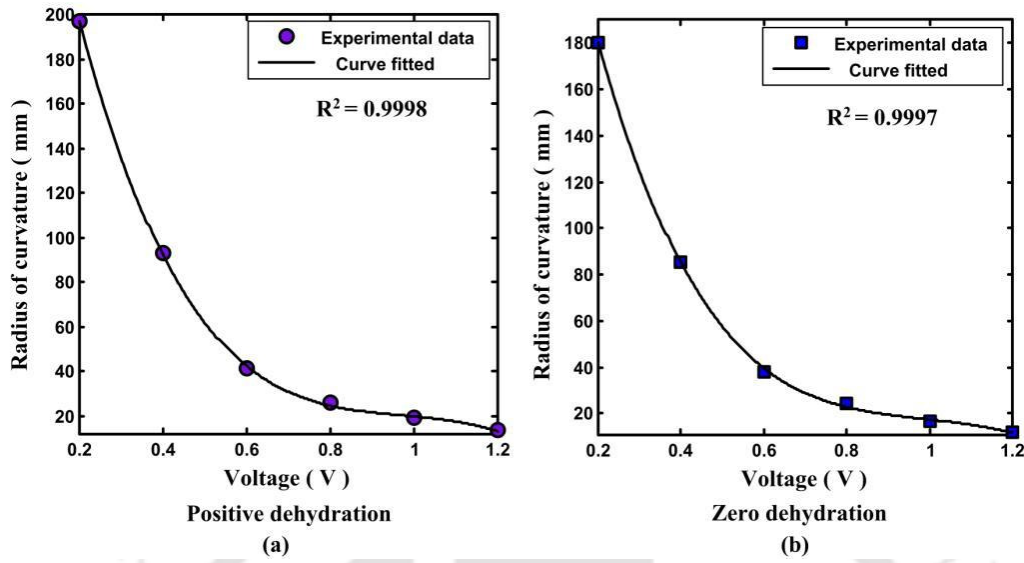


Figure 5.1 Radius of curvature at different input voltages (a) positive (b) zero dehydration.

5.2.2 Governing Equation of Motion

In this section the governing equation of motion of a silver-electroded IPMC having length(l), area of cross-section(A), modulus of elasticity(E), density(ρ), and area moment of inertia(I) is developed using d'Alembert's Principle. Figure 5.2 shows the schematic diagram of the IPMC actuator in a fixed-free configuration subjected to an alternating electric potential at its fixed end across its thickness. Mathematical modelling is done by incorporating the tip force generated due to electric potential. The actuator is modelled as uniform flexible distributed parameter system following the Euler-Bernoulli beam theory as shown in Figure 5.2. For an alternating electric potential, the equivalent tip force is assumed to be $F \cos \Omega t$ as shown in Figure 5.2. This equivalent force (F) for 'positive' and 'zero' dehydration condition can be calculated by substituting the expression for F_d and F_0 from equation 5.3(a) and 5.3(b) respectively.

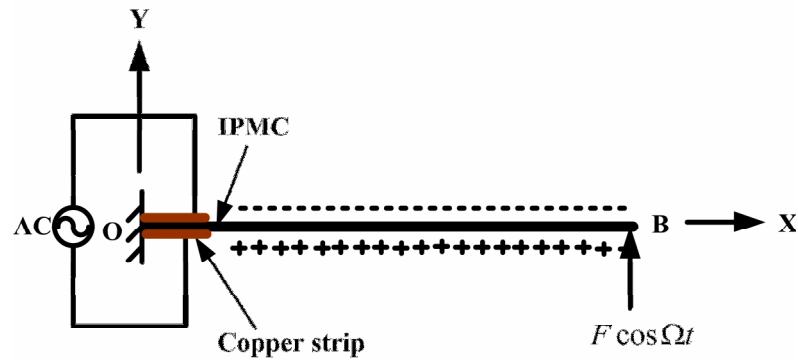


Figure 5.2 Schematic diagram: undeformed condition of IPMC actuator.

Figure 5.3 shows the schematic diagram of bending configuration of IPMC actuator subjected to alternating electric potential at the fixed end across its thickness. The actuator is operated by applying electric potential at the fixed end and allows it to bend at the free end. As the cross-section of the beam is assumed to be very small in comparison to the length of the IPMC, the beam is assumed to undergo pure bending and hence shear deformation and rotary inertia effects if any, are neglected in the mathematical modelling.

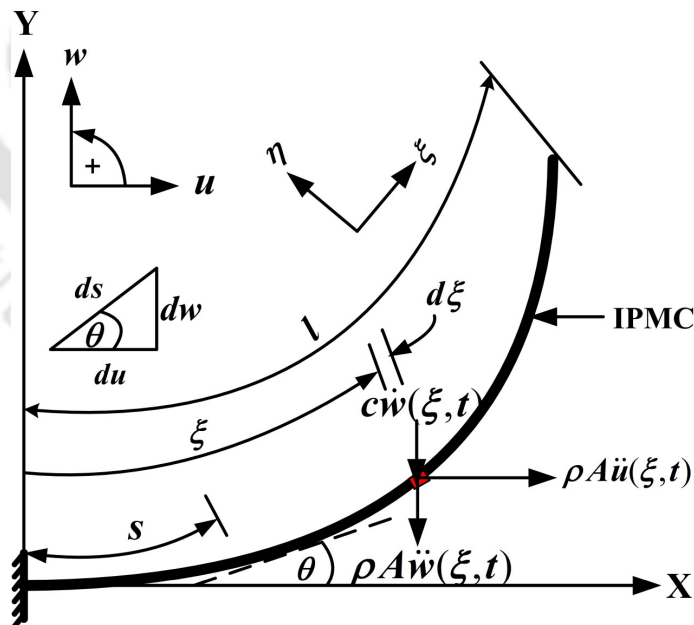


Figure 5.3 Bending configuration of an IPMC actuator for an input voltage V .

The bending moment of the link $M(s)$ at a distance s from the fixed end along the length of the link, as shown in Figure 5.3 can be expressed as [Zavodney and Nayfeh (1989), Cuvalci (2000)]:

$$M(s) \approx EI \left[w_{ss} + \frac{1}{2} w_s^2 w_{ss} \right] \quad (5.4)$$

where, w is the transverse displacement of the link. $(\cdot)_s$ is the first derivative with respect to curvature (s) along the link. Assuming the length of the link remains unchanged during bending i.e., inextensibility condition of the actuator, the longitudinal displacement (X-direction) $u(\xi, t)$ and the transverse displacement (Y-direction) $w(\xi, t)$ can be expressed as [Zavodney and Nayfeh (1989), Cuvalci (2000)]:

$$w_s^2 + (1 + u_s)^2 = 1 \quad \text{or,} \quad u(\xi, t) = \xi - \int_0^\xi (1 - w_\eta^2)^{\frac{1}{2}} d\eta \quad (5.5)$$

where, ξ, η are the integration variables. As shown in Figure 5.3, the inertia forces per unit length of the actuator are $\rho A \ddot{u}$ and $\rho A \ddot{w}$ in longitudinal and transverse directions, respectively. The bending moment due to the inertia forces in longitudinal ($M_{u\xi}(s)$) and transverse ($M_{w\xi}(s)$) directions at a distance ξ from the fixed end can be expressed as:

$$M_{u\xi} = - \int_s^l \rho A \ddot{u}(\xi, t) \int_s^\xi \sin \theta(\xi, t) d\eta d\xi \quad (5.6)$$

$$M_{w\xi} = - \int_s^l (\rho A \ddot{w}(\xi, t) + c \dot{w}(\xi, t)) \int_s^\xi \cos \theta(\xi, t) d\eta d\xi \quad (5.7)$$

The moment due to the external force generated by the electric field can be expressed as:

$$M_{f\xi} = \int_s^l \delta(s-l) F \cos(\Omega t) \int_s^\xi \cos \theta(\xi, t) d\eta d\xi \quad (5.8)$$

where, Ω is the frequency of the applied electric potential and δ is the Dirac delta function. Hence, applying d' Alembert's principle, one can write

$$M(s) = M_{u\xi} + M_{w\xi} + M_{f\xi} \quad (5.9)$$

Substituting equations (5.4 to 5.8) into (5.9) and differentiating the resulting equation twice with respect to s , using the Leibniz's rule and applying the binomial expansion, the equation of motion can be obtained as:

$$\begin{aligned}
 & EI \left(w_{ssss} + \frac{1}{2} w_s^2 w_{ssss} + 3w_s w_{ss} w_{sss} + w_{ss}^3 \right) + \rho A w_s \left[\int_0^\xi (\dot{w}_s^2 + w_s \ddot{w}_s) d\eta \right. \\
 & \left. - w_{ss} \left[\int_s^l \rho A \int_0^\xi (\dot{w}_s^2 + w_s \ddot{w}_s) d\xi d\eta + (1 - \frac{1}{2} w_s^2) [\rho A \dot{w} + c\dot{w}] \right. \right. \\
 & \left. \left. + w_s w_{ss} \left[\int_s^l (\rho A \ddot{w} + c\dot{w}) d\xi \right] \right] = F \cos(\Omega t) \quad (5.10)
 \end{aligned}$$

The nonlinear term associated with forcing is neglected. Taking single mode approximation, generalized Galerkin's method is used to discretize the governing equation of motion (5.10) into temporal form by substituting:

$$w(s, t) = r\psi(s)q(t) \quad (5.11)$$

where, $r, \psi(s)$, and $q(t)$ are the scaling factor, admissible function and time modulation. The admissible function $\psi(s)$ is considered to be same as the eigenfunction of a cantilever link and can be expressed as:

$$\psi(s) = -\frac{(\cos \beta l + \cosh \beta l)}{(\sin \beta l + \sinh \beta l)} (\sin \beta s - \sinh \beta s) + (\cos \beta s - \cosh \beta s) \quad (5.12)$$

where, $\beta l = 1.8751$ for the first mode of vibration. The following non-dimensional parameters are introduced for further analysis.

$$\bar{w} = \frac{w}{l}, \bar{s} = \frac{s}{l}, \bar{\xi} = \frac{\xi}{l}, \bar{\eta} = \frac{\eta}{l}, \tau = \omega_1 t, \bar{\Omega} = \frac{\Omega}{\omega_1} \quad (5.13)$$

ω_1 is the fundamental frequency of the system. Substituting the non-dimensional parameters (equation (5.13)) and equation (5.11) into equation (5.10) and using generalized Galerkin's method, the resulting non-dimensional temporal equation of motion is derived as:

$$\begin{aligned}
& \rho A r l \omega_1^2 \int_0^1 \{ \psi(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} \ddot{q}(\tau) + \frac{E l r}{l^3} \int_0^1 \{ \psi^{iv}(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} q(\tau) \\
& + C r l \omega_1 \int_0^1 \{ \psi(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} \dot{q}(\tau) + \frac{E l r^3}{2 l^3} \int_0^1 \{ \psi'^2(\bar{s}) \psi^{iv}(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} q^3(\tau) \\
& + \frac{E l r^3}{l^3} \int_0^1 \{ \psi'''(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} q^3(\tau) + \frac{3 E l r^3}{l^3} \int_0^1 \{ \psi'(\bar{s}) \psi''(\bar{s}) \psi'''(\bar{s}) \psi_w(\bar{s}) d\bar{s} \} q^3(\tau) \\
& + \rho A r^3 l \omega_1^2 \int_0^1 \left[\left\{ \psi'(\bar{s}) \psi''(\bar{s}) \int_{\bar{s}}^1 \psi(\bar{s}) d\bar{\xi} \right\} \psi_w(\bar{s}) d\bar{s} \right] q^2(\tau) \ddot{q}(\tau) \\
& + \rho A r^3 l \omega_1^2 \int_0^1 \left[\left\{ \psi'(\bar{s}) \int_0^{\bar{\xi}} \psi'(\bar{s}) \psi'(\bar{s}) d\bar{\eta} \right\} \psi_w(\bar{s}) d\bar{s} \right] q^2(\tau) \dot{q}(\tau) \\
& - \rho A r^3 l \omega_1^2 \int_0^1 \left[\left\{ \psi''(\bar{s}) \int_{\bar{s}}^1 \int_0^{\bar{\xi}} \psi'(\bar{s}) \psi'(\bar{s}) d\bar{\xi} d\bar{\eta} \right\} \psi_w(\bar{s}) d\bar{s} \right] q^2(\tau) \dot{q}(\tau) \\
& - \frac{\rho A r^3 l \omega_1^2}{2} \int_0^1 \left[\left\{ \psi(\bar{s}) \psi'^2(\bar{s}) \right\} \psi_w(\bar{s}) d\bar{s} \right] q^2(\tau) \ddot{q}(\tau) \\
& + C r^3 l \omega_1 \int_0^1 \left\{ \psi'(\bar{s}) \psi''(\bar{s}) \int_{\bar{s}}^1 (\psi(\bar{s}) d\bar{\xi}) \psi_w(\bar{s}) d\bar{s} \right\} q^2(\tau) \dot{q}(\tau) \\
& - \frac{C r^3 l \omega_1}{2} \int_0^1 \left\{ \psi(\bar{s}) \psi'^2(\bar{s}) \psi_w(\bar{s}) d\bar{s} \right\} q^2(\tau) \dot{q}(\tau) \\
& + \rho A r^3 l \omega_1^2 \int_0^1 \left[\left\{ \psi'(\bar{s}) \int_0^{\bar{\xi}} \psi'^2(\bar{s}) d\bar{\eta} \right\} \psi_w(\bar{s}) d\bar{s} \right] q(\tau) \dot{q}^2(\tau) \\
& - \rho A r^3 l \omega_1^2 \int_0^1 \left[\left\{ \psi''(\bar{s}) \int_{\bar{s}}^1 \int_0^{\bar{\xi}} \psi'^2(\bar{s}) d\bar{\xi} d\bar{\eta} \right\} \psi_w(\bar{s}) d\bar{s} \right] q(\tau) \dot{q}^2(\tau) \\
& = F \cos \bar{\Omega} \tau \int_0^1 \psi_w(\bar{s}) d\bar{s} \tag{5.14}
\end{aligned}$$

Simplifying the equation (5.14) and introducing the book keeping parameter, $\varepsilon (\ll 1)$ the resulting non-dimensional temporal equation of motion is obtained as:

$$\ddot{q} + q + 2\varepsilon \zeta \dot{q} + \varepsilon \alpha_1 q^3 + \varepsilon \alpha_2 q^2 \ddot{q} + \varepsilon \alpha_3 q^2 \dot{q} + \varepsilon \alpha_4 q \dot{q}^2 = \varepsilon F_n \cos(\bar{\Omega} \tau) \tag{5.15}$$

The expressions for the coefficients $(\omega_n, \zeta, \alpha_1, \alpha_2, \alpha_3, \alpha_4, F_n)$ are given below.

The system fundamental frequency, $\omega_1 = \sqrt{\frac{EIh_{11}}{\rho AL^4 h_1}}$

Damping ratio, $\zeta = \frac{ch_6}{2\rho A\omega_1^2 h_1}$

Coefficient of the non-linear geometric term (q^3), $\alpha_1 = \frac{EI r^2 \left(\frac{h_{12}}{2} + h_{13} + 3h_{14} \right)}{\varepsilon \rho AL^4 \omega_1^2 h_1}$

Coefficient of the non-linear inertia term ($q^2 \ddot{q}$), $\alpha_2 = \frac{r^2 \left(h_2 + h_3 - h_4 - \frac{h_5}{2} \right)}{\varepsilon h_1}$

Coefficient of the non-linear inertia term ($q^2 \dot{q}$), $\alpha_3 = \frac{cr^2 \left(h_7 - \frac{h_8}{2} \right)}{\varepsilon \rho A \omega_1 h_1}$

Coefficient of the non-linear inertia term ($q \dot{q}^2$), $\alpha_4 = \frac{r^2 (h_9 - h_{10})}{\varepsilon h_1}$

Coefficient of the linear force term, $F_n = \frac{F h_{15}}{\varepsilon \rho A r L \omega_1^2 h_1}$

The expression for $h_1, h_2, h_3, \dots, h_{15}$ are given below:

$$h_1 = \int_0^1 \psi(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_2 = \int_0^1 \psi'(\bar{s}) \int_0^{\bar{\xi}} \left(\{\psi'(\bar{s})\}^2 d\bar{\eta} \right) \psi_w(\bar{s}) d\bar{s},$$

$$h_3 = \int_0^1 \psi'(\bar{s}) \psi''(\bar{s}) \int_{\bar{s}}^1 \left(\psi(\bar{s}) d\bar{\xi} \right) \psi_w(\bar{s}) d\bar{s},$$

$$h_4 = \int_0^1 \psi''(\bar{s}) \int_{\bar{s}}^1 \int_0^{\bar{\xi}} \left(\{\psi'(\bar{s})\}^2 d\bar{\xi} d\bar{\eta} \right) \psi_w(\bar{s}) d\bar{s},$$

$$h_5 = \int_0^1 \psi(\bar{s}) \psi'^2(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_6 = \int_0^1 \psi(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_7 = \int_0^1 \psi'(\bar{s}) \psi''(\bar{s}) \int_{\bar{s}}^1 (\psi(\bar{s}) d\bar{\xi}) \psi_w(\bar{s}) d\bar{s},$$

$$h_8 = \int_0^1 \psi(\bar{s}) \psi'^2(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_9 = \int_0^1 \psi'(\bar{s}) \int_0^{\bar{\xi}} \psi'^2(\bar{s}) d\bar{\eta} \psi_w(\bar{s}) d\bar{s},$$

$$h_{10} = \int_0^1 \psi''(\bar{s}) \int_{\bar{s}}^1 \int_0^{\bar{\xi}} (\psi'^2(\bar{s}) d\bar{\xi} d\bar{\eta}) \psi_w(\bar{s}) d\bar{s}$$

$$h_{11} = \int_0^1 \{\psi''(\bar{s})\}^2 d\bar{s},$$

$$h_{12} = \int_0^1 (\psi'(\bar{s}))^2 \psi^{iv}(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_{13} = \int_0^1 (\psi''(\bar{s}))^3 \psi_w(\bar{s}) d\bar{s}$$

$$h_{14} = \int_0^1 \psi'(\bar{s}) \psi''(\bar{s}) \psi'''(\bar{s}) \psi_w(\bar{s}) d\bar{s},$$

$$h_{15} = \int_0^1 \psi_w(\bar{s}) d\bar{s}$$

It is observed that numerical values of the coefficients of damping; forcing and nonlinear terms are one order less than the coefficients of the linear terms, e.g., $\ddot{q}$ and q , which have unit value in this case. Hence, to make all the coefficients $\alpha_1, \alpha_2, \alpha_3, \alpha_4$ of the order of unity, a book keeping parameter ($\varepsilon \ll 1$) is introduced in equation of motion (5.15). It is observed that the non-linear temporal equation (5.15) contains linear forcing term F_n , cubic geometric term $\alpha_1 q^3$, nonlinear inertia

term $(\alpha_2 q^2 \ddot{q} + \alpha_4 q \dot{q}^2)$, and nonlinear damping term $\alpha_3 q^2 \dot{q}$. By incorporating loss-factor into the model, the equation of motion for the single-link IPMC actuator can be modified as:

$$\ddot{q} + q + 2\varepsilon \zeta \dot{q} + \varepsilon \alpha_1 q^3 + \varepsilon \alpha_2 q^2 \ddot{q} + \varepsilon \alpha_3 q^2 \dot{q} + \varepsilon \alpha_4 q \dot{q}^2 = \varepsilon \frac{F_{n0}}{(1-\lambda)} \cos(\bar{\Omega} \tau) \quad (5.16)$$

For zero dehydration condition i.e., $\lambda = 0$ the equation (5.16) reduces to the temporal equation of motion (equation 5.15). It is assumed that during dehydration only the external forcing parameter changes. This observation later can be shown to be valid by comparing the experimental results.

5.3 Solution of Temporal Equation of Motion

As the closed form solution or exact solution of equations (5.15) or (5.16) is difficult to obtain, an approximate solution is obtained by using perturbation technique and thus, method of multiple-scales has been used to solve the equation of motion. In the following sections, solutions of the temporal equation of motion and numerical simulations have been carried out for the actuator taking into account of dehydration effect. Several perturbation methods i.e. method of multiple scales (MMS), averaging method (AM), Lindstedt-Poincaré method (LPM), and method of normal forms (MNF) can be used to solve the nonlinear equation. In the present work, method of multiple scales has been used to find the solution of equation (5.15). The method is straight forward, further knowledge about the solution before the analysis is not required and it is very efficient for a multi-degree-of-freedom (MDOF) nonlinear analysis. The response and stability criteria can be obtained from the same analysis. Following the standard procedure of Method of Multiple Scales [Nayfeh and Mook (1995), and Nayfeh and Balachandran (1995)], the displacement q is expressed in terms of different time scales (T_0, T_1) and a book keeping parameter ε as follows.

$$q(\tau; \varepsilon) = q_0(T_0, T_1) + \varepsilon q_1(T_0, T_1) + \dots \quad (5.17)$$

where, $T_0 = \tau$, $T_1 = \varepsilon \tau, \dots$

The first and second time derivatives are as:

$$\begin{aligned}\frac{d}{d\tau} &= \frac{dT_0}{d\tau} \frac{\partial}{\partial T_0} + \frac{dT_1}{d\tau} \frac{\partial}{\partial T_1} + \dots = D_0 + \varepsilon D_1 + \varepsilon^2 D_2 + \dots \\ \frac{d^2}{d\tau^2} &= D_0^2 + 2\varepsilon D_0 D_1 + \varepsilon^2 (D_1^2 + 2D_0 D_2) + \dots\end{aligned}\quad (5.18)$$

Substituting equations (5.17) and (5.18) into equation (5.15) and equating the coefficients of the power terms of ε , one can obtain the following expressions.

$$\text{Order } \varepsilon^0 : D_0^2 q_0 + q_0 = 0 \quad (5.19)$$

$$\begin{aligned}\text{Order } \varepsilon^1 : D_0^2 q_1 + q_1 &= F_{nd} \cos(\bar{\Omega} T_0) - 2D_0 D_1 q_0 - 2\xi D_0 q_0 - \alpha_1 q_0^3 \\ &\quad - \alpha_2 q_0^2 (D_0^2 q_0) - \alpha_3 q_0^2 (D_0^2 q_0) - \alpha_4 q_0 (D_0 q_0)^2\end{aligned}\quad (5.20)$$

The general solution of the equation (5.19) is of the form:

$$q_0 = A(T_1) e^{iT_0} + \bar{A}(T_1) e^{-iT_0} \quad (5.21)$$

Substituting the value of q_0 from equation (5.21) into equation (5.20), and expressing $\cos(\bar{\Omega} T_0)$ in exponential form:

$$\begin{aligned}D_0^2 q_1 + q_1 &= \left(-2D_1 iA - 2\xi iA - 3\alpha_1 A^2 \bar{A} + 3\alpha_2 A^2 \bar{A} - \alpha_3 iA^2 \bar{A} - \alpha_4 A^2 \bar{A} \right) e^{iT_0} \\ &\quad + \left(-\alpha_1 A^3 + \alpha_2 A^3 - \alpha_3 iA^3 + \alpha_4 A^3 \right) e^{3iT_0} + \frac{1}{2} F_{nd} e^{i\bar{\Omega} T_0} + cc\end{aligned}\quad (5.22)$$

where cc stands for the complex conjugate of the preceding terms. It is observed that any solution of equation (5.22) contains both secular and mixed-secular terms when frequency of the electric potential ($\bar{\Omega}$) is nearly equal to 1. This gives rise to simple resonance case and is discussed in the following section. For the simple resonance case, a detuning parameter σ is used to express the nearness of $\bar{\Omega}$ to 1 as:

$$\bar{\Omega} = (1 + \varepsilon \sigma), \text{ and } \sigma = O(1) \quad (5.23)$$

Eliminating the secular or small divisor terms of equation (5.22), yields,

$$-2D_1 iA - 2\xi iA - 3\alpha_1 A^2 \bar{A} + 3\alpha_2 A^2 \bar{A} - \alpha_3 iA^2 \bar{A} - \alpha_4 A^2 \bar{A} + \frac{1}{2} F_{nd} e^{i\sigma T_1} = 0 \quad (5.24)$$

Substituting, $A = \frac{1}{2}ae^{i\beta}$ where a and β are real numbers, in equation (5.24) and separating the real and imaginary parts of the resulting equation the following expressions are obtained.

$$a\beta' = \frac{3}{8}\alpha_1 a^3 - \frac{3}{8}\alpha_2 a^3 + \frac{1}{8}\alpha_4 a^3 - \frac{1}{2}F_{nd} \cos(\sigma T_1 - \beta) \quad (5.25)$$

$$a' = -\xi a - \frac{1}{8}\alpha_3 a^3 + \frac{1}{2}F_{nd} \sin(\sigma T_1 - \beta) \quad (5.26)$$

Introducing $\gamma = \sigma T_1 - \beta$ into the equations (5.25) and (5.26), yields

$$a\gamma' = a\sigma - \frac{3}{8}\alpha_1 a^3 + \frac{3}{8}\alpha_2 a^3 - \frac{1}{8}\alpha_4 a^3 + \frac{1}{2}F_{nd} \cos(\gamma) \quad (5.27)$$

$$a' = -\xi a - \frac{1}{8}\alpha_3 a^3 + \frac{1}{2}F_{nd} \sin(\gamma) \quad (5.28)$$

For steady-state response (a_0, γ_0) , $a' = \gamma' = 0$. Hence equations (5.27) and (5.28) are reduced to,

$$-a\sigma + \frac{3}{8}\alpha_1 a^3 - \frac{3}{8}\alpha_2 a^3 + \frac{1}{8}\alpha_4 a^3 = \frac{1}{2}F_{nd} \cos(\gamma) \quad (5.29)$$

$$\xi a + \frac{1}{8}\alpha_3 a^3 = \frac{1}{2}F_{nd} \sin(\gamma) \quad (5.30)$$

Thus, the frequency response of the system can be obtained by solving equations (5.29, 5.30) simultaneously. Eliminating γ from equations (5.29, 5.30), one may obtain the relation between detuning parameter (σ) and amplitude (a) with positive dehydration as:

$$\sigma = Ka^2 \pm \sqrt{\left(\frac{F_{nd}}{2a}\right)^2 - \left(\xi + \frac{\alpha_3 a^2}{8}\right)^2} \quad (5.31)$$

Similarly, in case of zero dehydration, the Equation (5.31) is modified to:

$$\sigma = Ka^2 \pm \sqrt{\left(\frac{F_{n0}}{2a}\right)^2 - \left(\xi + \frac{\alpha_3 a^2}{8}\right)^2} \quad (5.32)$$

where, $K = \left\{ \frac{3\alpha_1}{8} - \frac{3\alpha_2}{8} + \frac{\alpha_4}{8} \right\}$. Equation (5.31) is an implicit form for amplitude of

the response as a function of excitation frequency, damping ratio ξ and other system

parameters. It is observed that, equation (5.31) does not possess any trivial state response, i.e., $a=0$ is not a solution. The response of the system can be obtained either by using the equation (5.31) or numerically solving the reduced equations (5.29) and (5.30), simultaneously. Also, the temporal equation of motion (5.15) can be numerically solved to get the response for a particular set of system parameters. The stability of the steady-state response (a_0, γ_0) can be determined, by perturbing the equations (5.27) and (5.28), by substituting $a = a_0 + \Delta a$ and $\gamma = \gamma_0 + \Delta \gamma$, and then investigating the eigenvalues of the resulting Jacobian matrix (J) which is obtained as:

$$J = \begin{bmatrix} -\zeta - \frac{3\alpha_3 a_0^2}{8} & -a_0 \sigma + \frac{3\alpha_1 a_0^3}{8} - \frac{3\alpha_2 a_0^3}{8} + \frac{\alpha_4 a_0^3}{8} \\ \left(\sigma - \frac{9\alpha_1 a_0^2}{8} + \frac{9\alpha_2 a_0^2}{8} - \frac{3\alpha_4 a_0^2}{8} \right) \frac{1}{a_0} & -\zeta - \frac{\alpha_3 a_0^2}{8} \end{bmatrix} \quad (5.33)$$

In this simple resonance condition, the system with steady-state motion will be stable, if and only if all the real parts of the eigenvalues are negative. From equation (5.21), the first-order non-trivial approximate steady-state solution is given by:

$$\begin{aligned} q = q_0 &= A e^{iT_0} + A e^{-iT_0} = \frac{1}{2} a e^{i\beta} e^{iT_0} + \frac{1}{2} a e^{-i\beta} e^{-iT_0} \\ &= \frac{1}{2} a \left(e^{i(\sigma T_1 - \gamma + T_0)} + e^{-i(\sigma T_1 - \gamma + T_0)} \right) = \frac{1}{2} a \left(e^{i\{(1+\varepsilon\sigma)T_0 - \gamma\}} + e^{-i\{(1+\varepsilon\sigma)T_0 - \gamma\}} \right) \end{aligned} \quad (5.34)$$

Hence,

$$q = a \cos(\bar{\Omega} \tau - \gamma) \quad (5.35)$$

For the system with zero dehydration, in the above equation F_{nd} can be replaced by F_{n0} .

5.4 Results and Discussions

In this section theoretical results are obtained by numerically solving equation 5.31, the frequency response curves have been obtained for different system parameters. This result has been verified by solving the temporal equation of motion. Further, results have been validated by conducting experiment are discussed in section 5.4.2.

5.4.1 Numerical Results

An IPMC actuator is numerically studied and non-linear response is investigated. A silver electroded IPMC actuator of length (L) = 0.02 m, width (w_d) = 0.005 m and thickness (h) = 0.0002 m has been considered for the numerical analysis. The density (ρ) and Young's modulus (E) are taken 2125 kg/m³ and 82 MPa (same as previous chapter), respectively. Both book-keeping parameter (ε) and scaling factor (r) are taken as 0.1 for simulation. All the results in this section are shown considering dehydration. Variation of non-dimensional force term with input voltages is given in Table 5.1.

Table 5.1 Variation of non-dimensional force term with voltage.

Voltage (V)	F_{nd}	F_{n0}
	(positive dehydration)	(zero dehydration)
0.2	0.6516	0.7087
0.4	1.4092	1.5226
0.6	3.1714	3.3333
0.8	5.8895	6.1022
1.0	8.2612	8.8769
1.2	15.2365	16.2023

Equation (5.31) is solved and the results are verified with the results by solving the temporal equation of motion (5.15). The multiple equilibrium points obtained in the frequency response curve is validated by plotting the basin of attraction. As evident, due to the presence of various non-linear terms in the temporal equation of motion (5.15), the system yields a typical non-linear behavior. For simple resonance case, the frequency response curve is obtained by using equation (5.31) which is shown in Figures 5.4-5.7 for different system parameters. While the solid line represents the stable equilibrium points the dashed line represents the unstable state.

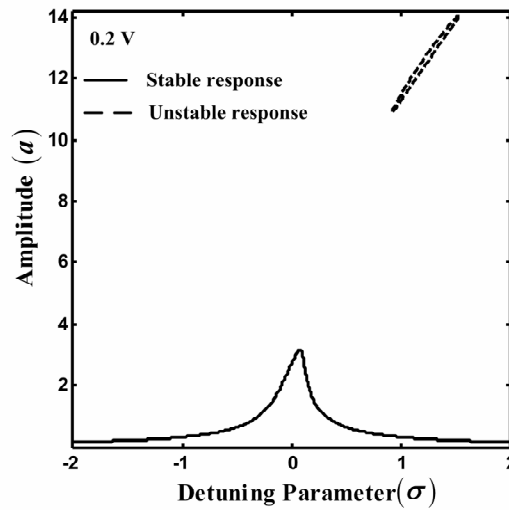


Figure 5.4 Frequency response curve for input voltage of 0.2V.

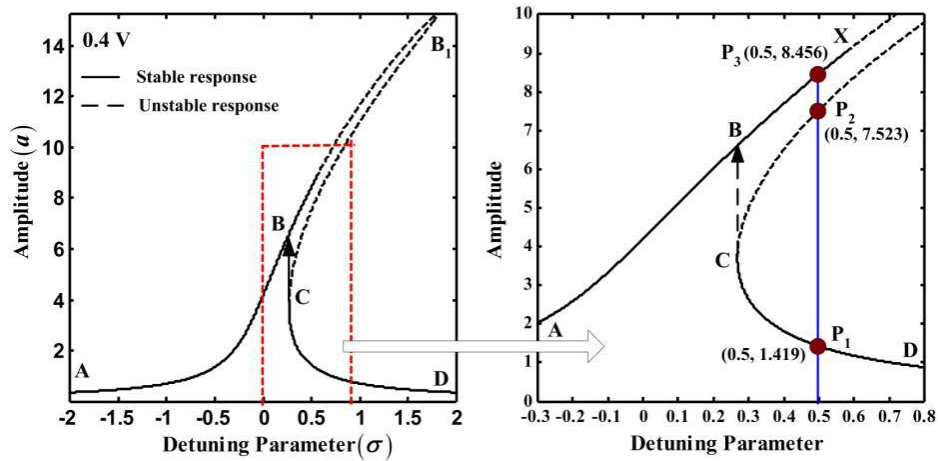


Figure 5.5 Frequency response curve for input voltage of 0.4V.

It is observed from the results that the system does not possess any trivial state response. It is thus, anticipated that the actuator always vibrates about its equilibrium position with an amplitude equal to the non-trivial response as shown in Figure 5.4, 5.5, 5.6 and 5.7. In Figure 5.4 the frequency response curve is plotted for 0.2V and it is observed an isolated unstable regime in addition to a stable branch. With increase in voltage further these two branches merge to form a frequency response curve similar to Figure 5.5-5.7. It is clearly observed that frequency range

XC (Figure 5.5) demarcates the unstable response of the system while frequency range AX and CD shows the stable regime. When the actuator is activated (e.g. at point A), with increase in frequency of excitation, the amplitude of response increases and it reaches a critical value at point X. At this point, further increase in frequency, leads the system to unstable. Further, increase in frequency drives to the critical point C, beyond which the system again exhibits stable condition. It is further observed that, frequency beyond point C, the amplitude decreases to point D. This observation is consistent with the results presented by Nemat-Nasser and Wu (2006) and Cilingir and Papila (2010). If the system is driven by a voltage with frequency in between point X and C the response will be thus unstable, and may results in failure in this range. Hence, to operate the system with moderate amplitude of response, either the applied voltage should be lowered or the operating frequency should be kept well below the bifurcation point X or above of the bifurcation point C. It is also observed that as the input voltage increases the response amplitude also increases as demonstrated in Figure 5.5, Figure 5.6 and Figure 5.7(a,b).

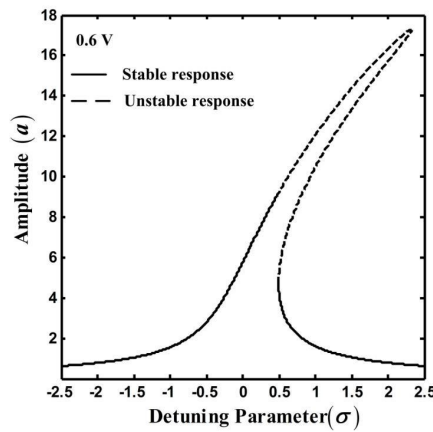


Figure 5.6 Frequency response curve for input voltage of 0.6V.

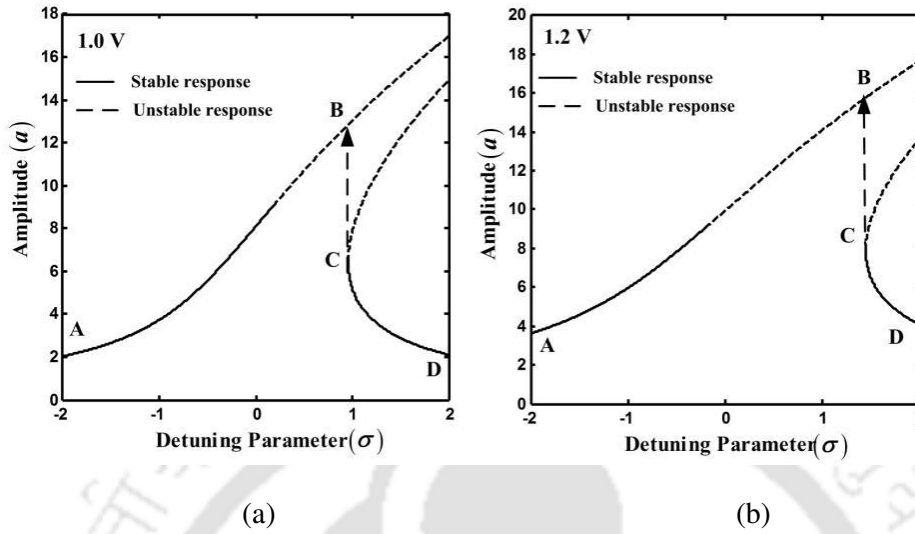


Figure 5.7 Frequency response curve for input voltage of (a) 1.0V (b) 1.2V.

Similarly, when the excitation frequency is swept down, i.e., from point D, with decrease in frequency the response amplitude increases and it reaches a critical value at point C, which is a saddle node bifurcation point. At this point with further increase in frequency, the system experiences a jump up phenomenon i.e. a sudden jump from point C to point B which leads to a sudden increase of amplitude and may lead to the system failure. Table 5.2 lists the variation of bifurcation points C and B, and the jump length for different input voltages. It is observed that with increase in the value of excitation voltage, the jump length increases. Table 5.3 lists the variation of bifurcation points C and X with input voltage. Further, it is clearly observed that the unstable zone, CX increases with increase of input voltage.

As shown in Figure 5.5, due to the presence of bi-stable region between the frequency ranges B-B₁, the initial conditions in this region plays an important role to find the appropriate system response. Hence, in order to know the influence of initial condition, basin of attraction is plotted in the ' $a - \gamma$ ' plane corresponding to $\sigma = 0.5$ for 0.4V as shown in Figure 5.8. It clearly shows two stable solutions corresponding to point (P₁, P₃) and one unstable solution corresponding to point (P₂) as marked in Figure 5.5. Figure 5.9 shows the time response and phase portrait, obtained by solving the equation (5.27) and (5.28) for applied electric potential of 0.4V. These results are in good agreement with the results presented in Figure 5.5 at point P₁.

Table 5.2 Variation of bifurcation points C and B with excitation voltage.

Voltage (V)	At critical point C		At critical point B		Jump length
	Detuning parameter	Response amplitude	Detuning parameter	Response amplitude	
	(σ)	(a)	(σ)	(a)	
0.4	0.2681	3.779	0.2681	6.579	2.8
0.6	0.4955	4.747	0.4955	9.21	4.463
0.8	0.7616	5.744	0.7616	11.48	5.736
1.0	0.9589	6.537	0.9589	12.83	6.293
1.2	1.447	7.989	1.447	15.75	7.761

Table 5.3 Variation of the critical points X and C with excitation voltage.

Voltage (V)	Detuning parameter (σ)	
	At critical point X	At critical point C
0.4	0.5786	0.2681
0.6	0.4701	0.4955
0.8	0.3126	0.7616
1.0	0.1899	0.9589
1.2	-0.2079	1.447

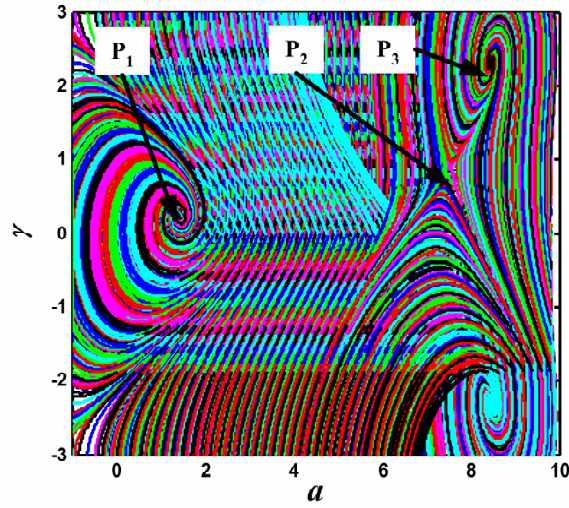
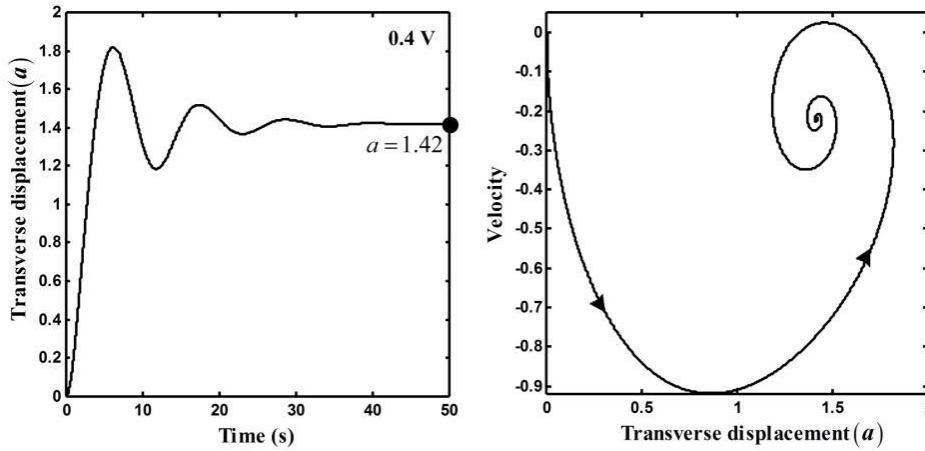
Figure 5.8 Basin of attraction for $\sigma = 0.5$, 0.4 V, Figure 5.5.Figure 5.9 Time response and phase portrait at point, $\sigma = 0.5$, 0.4 V, Figure 5.5.

Figure 5.10 shows the vibration response and phase portrait at a point, where $\sigma = 0.5$ as shown in Figure 5.5. The results are obtained by solving the temporal equation of motion (5.15 (a)). It is observed that, the transient response (Figure 5.10 (i),(ii)) of the system produces a beating type phenomenon, and the steady-state response of the system is periodic (Figure 5.10 (iii),(iv)). It is observed that the steady-state response obtained is in good agreement with the results obtained by using the method of multiple scales as shown in Figure 5.5. Figure 5.11 shows the time response for $\sigma = 0.5$, 1.0 V, which is found to be unstable by solving the

temporal equation of motion and can be well verified with the results presented in Figure 5.7 (a).

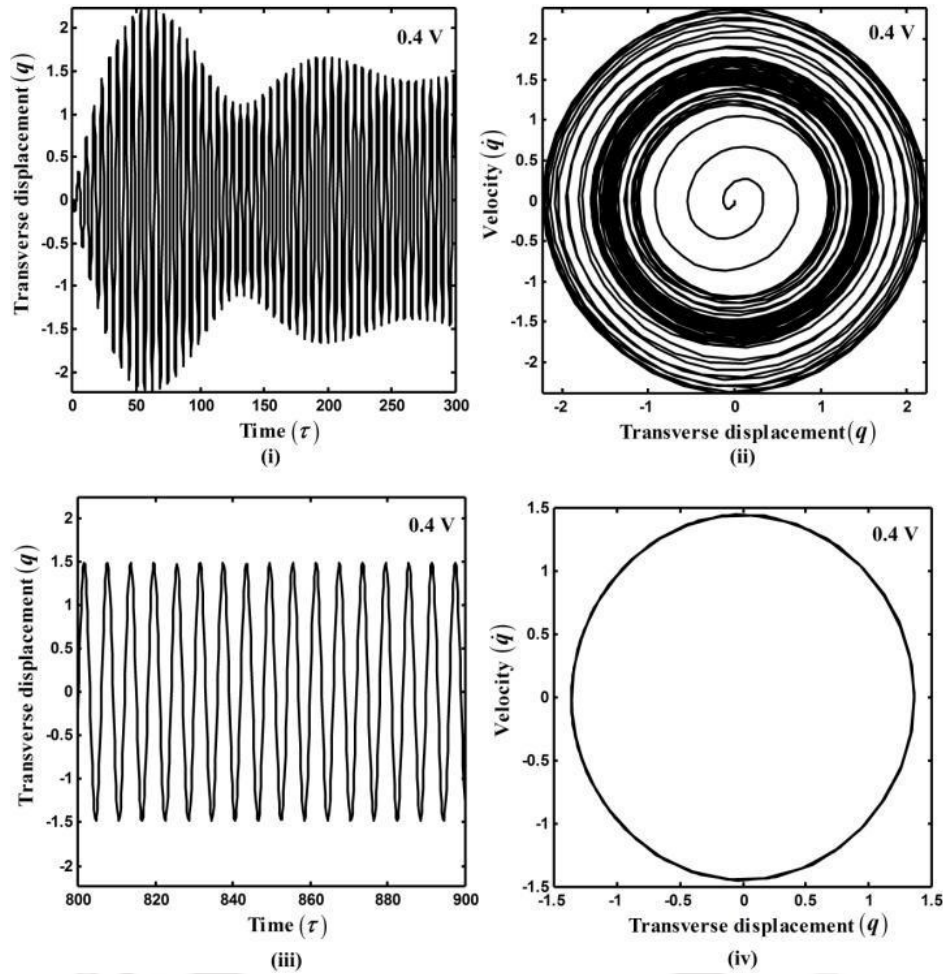


Figure 5.10 Vibration response and phase portrait for $\sigma = 0.5$, obtained by solving temporal equation of motion, (Key as in Figure 5.5), where, (i), (ii) transient response and (iii), (iv) steady-state response.

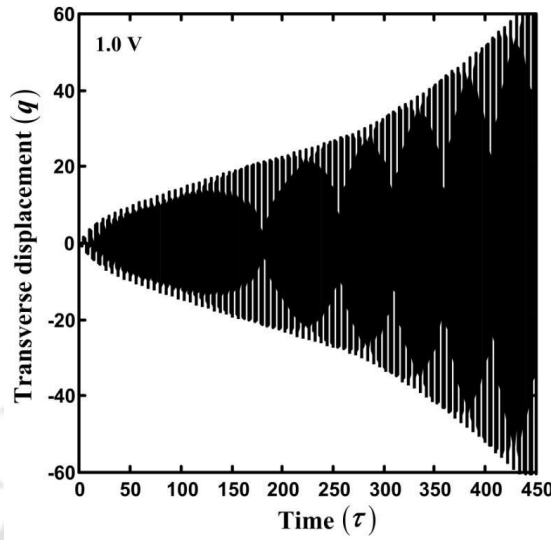


Figure 5.11 Unstable response for $\sigma = 0.5$, obtained by solving temporal equation of motion for an input voltage 1.0 V (Key as in Figure 5.6 (a)).

5.4.2 Experimental Results

An IPMC actuator of size $20 \times 5 \times 0.2$ (mm³) is prepared and tested in fixed-free configuration at different applied frequencies under sinusoidal input voltage. Transient and steady-state vibration characteristics are studied and compared with the theoretical results. The photograph of the experimental setup is shown in Figure 5.12. One end of the control module is connected to the Rotational Laser Vibrometer (RLV-5500; make: Polytec, GmbH) and other end to the computer where the data are processed using Bruel and Kjaer PULSE Lab Shop, Version 13.1.0.246 software. The function generator (Sony/Tektronix AFG 320) is used to apply AC voltage at the fixed end of the sample while Laser is directed on tip of the free end.

Before the experiment, IPMC sample is boiled in deionized water for 30 min and kept within deionized water at room temperature for 24h to ensure full hydration. The entrapped surface moistures are removed by keeping the actuator in open environment for 5 min prior to the starting of the experiment. Voltage is applied for 30s and subsequently response of the free end is measured using Rotational Laser Vibrometer. All the experiments are conducted at room temperature. The experimental results are presented in Figures 5.13-5.16.

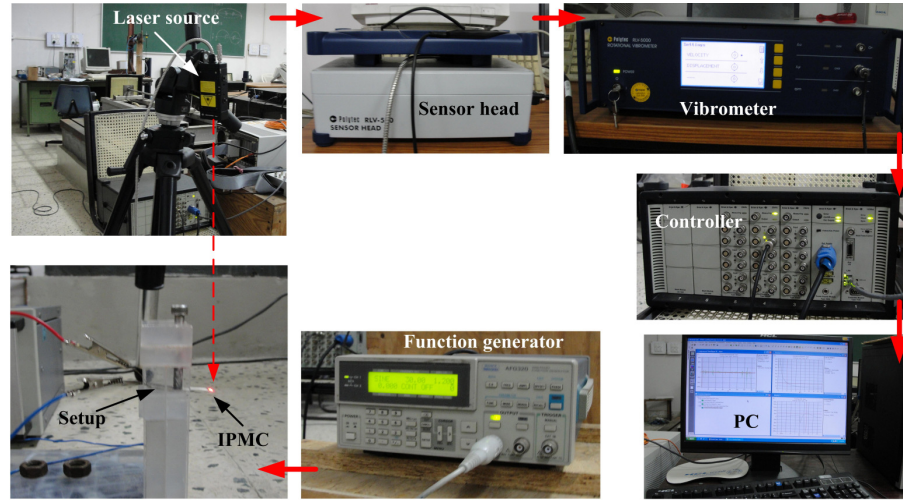


Figure 5.12 Photograph of the experimental setup for nonlinear vibration response analysis.

Figures 5.13 (a) and (b) demonstrate the vibration response of the IPMC actuator for an applied voltage of 0.4V at different frequency. It is observed that as the input frequency increases from 12.5 Hz (corresponds to $\sigma = -1$) to 16.5 Hz ($\sigma = 2$) the vibration amplitude decreases. For an input voltage of 0.4V, corresponding to a frequency of 12.5 Hz ($\sigma = -1$) the nontrivial amplitude of time modulation of steady-state response (a) is found to be around 0.004 mm. As shown in Figure 5.5, the transverse displacement (steady-state response amplitude) corresponding to $\sigma = -1$ is found to be $w(s, t) = \frac{r}{l} \psi(\bar{s}) q(\bar{t}) = 0.00564$ mm. Hence, the response obtained from the perturbation analysis is found to be in good agreement with the experimental value. Similarly for the point D ($\sigma = 2$) in Figure 5.5, the amplitude is found to be 0.00352 mm. While for the corresponding point D in Figure 5.13(b), the transverse displacement is found to be around 0.003 mm, which is in good agreement with the numerical results as shown in Figure 5.5. Figure 5.14 (a) and (b) show the vibration response at different input voltages with constant input frequency of 15.251 Hz ($\sigma = 1$). It is observed that as the input voltage increases the amplitude of the vibration response also increases, which also validates the simulation results shown in Figure 5.5 and Figure 5.6.

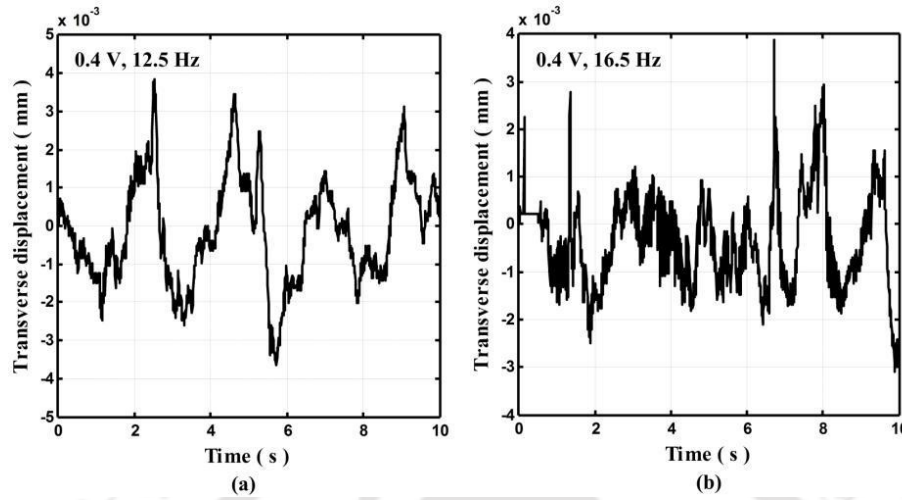


Figure 5.13 Experimental vibration response at 0.4V (a) 12.5 Hz (b) 16.5 Hz.

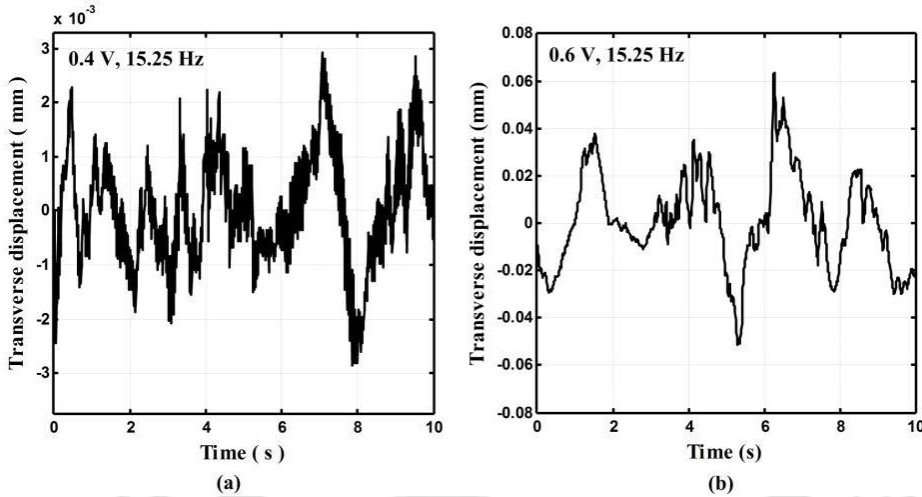


Figure 5.14 Experimental vibration response at 15.25 Hz (a) 0.4V (b) 0.6V.

Figure 5.15 (a) shows the unstable vibration response of the actuator at 1.0V with an applied frequency of 14.5 Hz, corresponds to $\sigma = 0.5$. It is observed that when the voltage is applied the amplitude increases gradually. The results also validate the simulation results as shown in Figure 5.7(a) and Figure 5.11. Figure 5.15(b) shows the steady-state response (after the voltage is withdrawn) at 1.0V with an applied frequency 8.0 Hz corresponding to $\sigma = -4$. Figure 5.16 demonstrates the FFT of the vibration response as shown in Figure 5.13(a). It is observed that band-width of vibration response remains within 2-30 Hz.

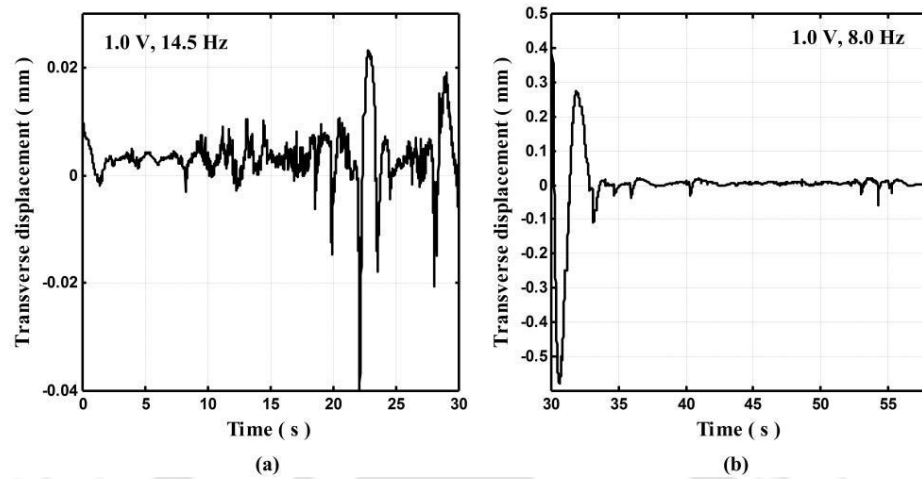


Figure 5.15 Experimental vibration response at 1.0V (a) Unstable (b) Stable.

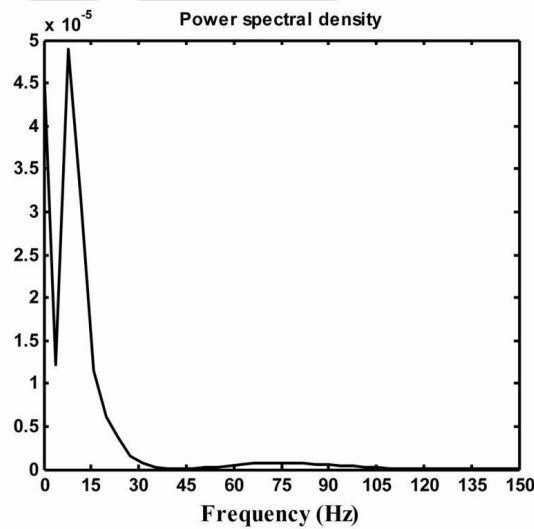


Figure 5.16 FFT of the vibration response at 0.4 V, 12.5 Hz.

5.5 Effect of Dehydration on the Vibration Response

Effect of dehydration on the nonlinear vibration response, has also been assessed by numerically solving the equations (5.31) and (5.32) the frequency response curve has been obtained for different system parameters. This result has been verified by solving the temporal equation of motion. Further some results have been validated by conducting experiment are discussed in section 5.5.2.

Figure 1 consists of two plots, (a) and (b), showing the amplitude response of the system.

Plot (a) shows Amplitude (a) versus Detuning Parameter (σ) for a voltage of 0.4 V. The x-axis ranges from -2 to 2, and the y-axis ranges from 0 to 14. A solid line represents the stable response, and a dashed line represents the unstable response. The curves are shown for positive dehydration (black) and zero dehydration (red). A vertical line at $\sigma = 1$ indicates a transition point. A legend at the bottom indicates: — Positive dehydration, — Zero dehydration.

Plot (b) shows Amplitude (a) versus Detuning Parameter (σ) for positive dehydration (black) and zero dehydration (red). The x-axis ranges from -0.1 to 0.8, and the y-axis ranges from 0 to 10. The curves show a stable response (solid line) and an unstable response (dashed line). Points A, B, C, D, P₁, P₂, P₃, and X are marked on the curves. A vertical line at $\sigma = 1$ indicates a transition point. A legend at the bottom indicates: — Positive dehydration, — Zero dehydration.

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Table 5.4 Variation of the bifurcation points C and B with excitation voltage for zero dehydration.

Voltage (V)	At critical point C		At critical point B		Jump length
	Detuning parameter	Response amplitude	Detuning parameter	Response amplitude	
	(σ)	(a)	(σ)	(a)	
0.4	0.2857	3.838	0.2857	6.839	3.001
0.6	0.5133	4.823	0.5133	9.372	4.549
0.8	0.7803	5.81	0.7803	11.58	5.77
1.0	1.007	6.671	1.007	13.14	6.469
1.2	1.508	8.034	1.508	16.07	8.036

Figure 5.18 shows the time response and phase portrait, for 'positive' and 'zero' dehydration condition obtained by solving the equation (5.27) and (5.28) for an input of 0.4V. It is clearly observed that the response amplitude increases in case of zero dehydration condition compared to positive dehydration conditions.

Table 5.5 Variation of the critical points X and C with excitation voltage for zero dehydration.

Voltage (V)	Detuning parameter (σ)	
	At critical point X	At critical point C
0.4	0.5694	0.2857
0.6	0.4578	0.5133
0.8	0.3075	0.7803
1.0	0.1436	1.007
1.2	-0.1706	1.508

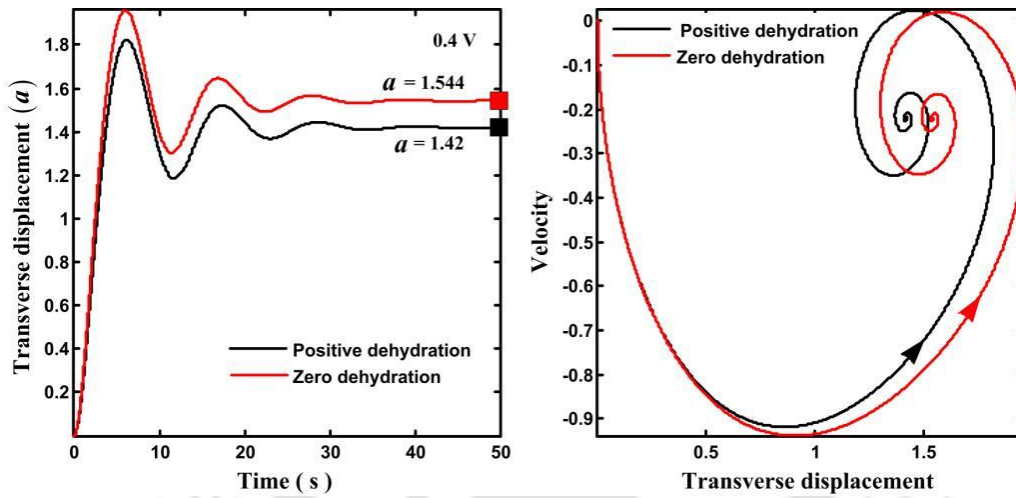


Figure 5.18 Effect of dehydration on Time response and Phase portrait curve at point, $\sigma = 0.5$, $0.4V$, (Figure 5.5 & Figure 5.16).

5.5.2 Experimental Results

Before the experiment, the same IPMC sample of size $20 \times 5 \times 0.2$ (mm^3) is kept in open environment for 5min, 10 min, 15 min, 30 min and 60 min, successively prior to the next experiment to ensure adequate dehydration. Figure 5.19 (a) and (b) show the vibration response of the actuator for initial (fully hydrated) condition and after 5 min from the initial condition and it is observed that peak amplitude reduces considerably due to dehydration.

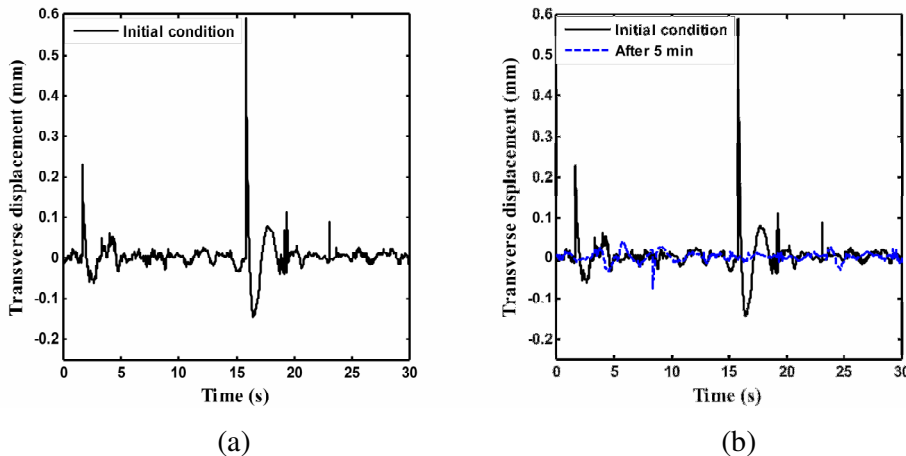


Figure 5.19 Experimental vibration response (a) initial condition (b) comparison between initial and after 5 min for an input of $0.2V$, 1Hz sine wave.

Figures 5.20 (a) and (b) show the vibration response of the actuator after 10 and 15 min; and 30 and 60 min, respectively from the starting of the experiment. It is observed that as the IPMC actuator gets dehydrated (dried) the vibration response remains almost same. Figure 5.21 shows unstable vibration response of the actuator for initial and after 10 min from the initial condition. It is observed that the amplitude of vibration decreases due to dehydration effect. The results are in good agreement with the simulated results, figure 5.6 (a) for $\sigma = 0.2$, which corresponds to 14.13 Hz.

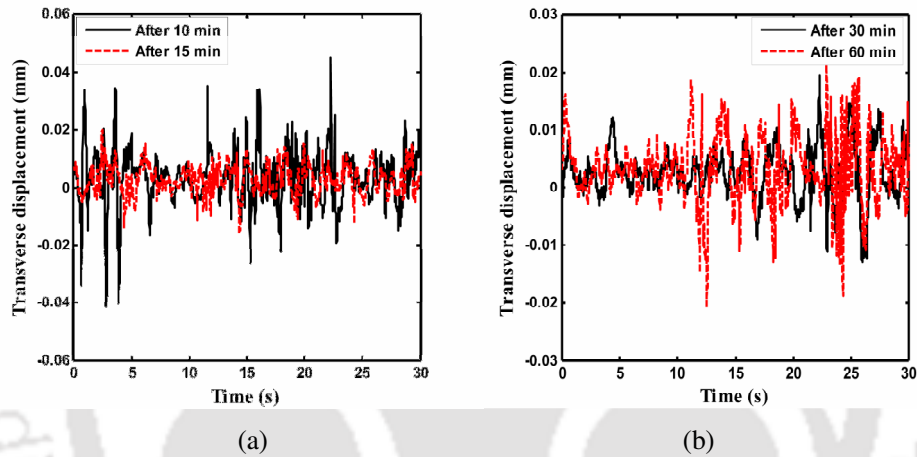


Figure 5.20 Experimental vibration responses (a) after 10 and 15 min (b) after 30 and 60 min for an input of 0.2V, 1Hz sine wave.

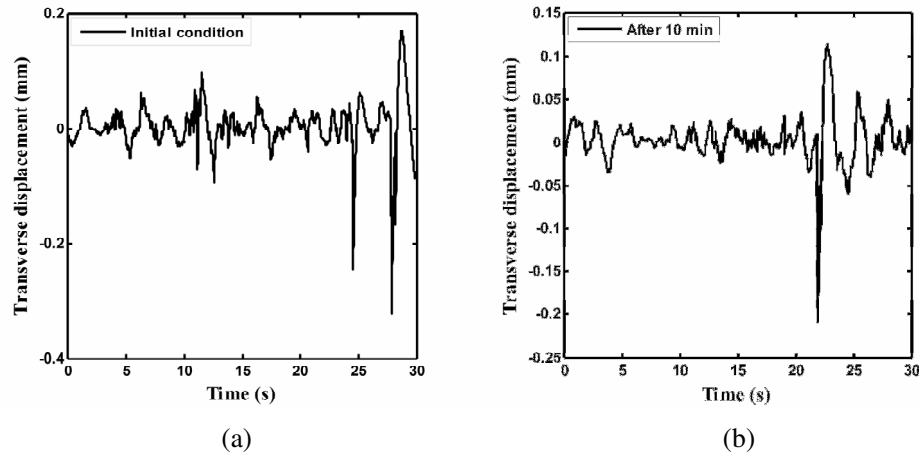


Figure 5.21 Experimental vibration response (a) initial condition (b) after 10 min for an input of 0.6V, 14Hz sine wave.

5.6 Summary

In this work, a closed form expression for the response amplitude and detuning parameter has been developed which can be used to obtain the nonlinear frequency response curve for any IPMC actuator. This simplified expression can be used for finding the response of the system instead of solving the temporal equation of motion which is time consuming and requires more memory space. The result obtained from perturbation analysis has been compared with those obtained by solving the temporal equation of motion and are found to be in good agreement. The numerical results are compared and verified by performing experiment. The precise predication of stable and unstable zone for an applied excitation voltage and frequency is of importance in many micromechanical systems utilizing IPMC as the actuator. Both theoretical and experimental investigations confirm that the amplitude of vibration response decreases with ‘positive’ dehydration. It is concluded that in micro-scale actuator, amplitude of vibration could increase considerably when a small change in excitation of amplitude occurs. Further, in micro-scale applications, the amplitude of vibration can be easily increased and it may bring the system into non-linear regime, and therefore, it is useful to utilize the non-linear analysis to predict the behavior of the system more accurately.

Application of Ag-IPMC as an Active Catheter System

6.1 Introduction

Active materials are nowadays utilized successfully for interventional diagnosis and therapy for their biocompatibility, suitability for smooth and safety operation. Active catheter is one such mechanism where active materials such as shape memory alloys (SMA), electroactive polymers (EAPs) and piezoelectric materials (PZT) are successfully used over the few years. Recently, there has been tremendous progress and demand for minimally invasive surgical (MIS) tools essential for medical diagnostics and treatment. Such tools sometimes use active catheter system which operates by giving force input and rely on the intrinsic mechanical property of long flexible catheter to transmit the motion to the distal tip. Drawback like hysteresis and recoiling results in poor controllability of these systems. Hence, there is deficiency in getting the desired accuracy and repeatability at the desired destinations. Shape memory alloy (SMA) actuators have been used for active catheters by utilizing their shape recovery effect with a heating above phase transformation temperature [Lim and Lee (1999); Park and Esashi (1999); Haga *et al.*, (2000); Mineta *et al.*, (2001); Tung *et al.*, (2007); Mineta *et al.*, (2011)]. Literatures showed that to fabricating multi-dimensional bending mechanism for active catheter, complex assembly processes, such as accurate alignment and gluing of multiple SMA actuators on a cylindrical surface of the catheter were necessary [Lim and Lee (1999); Haga *et al.*, (2000); Mineta *et al.*, (2001)]. Eventhough catheters made up of SMA are able to provide a large bending, their slow response and high operating temperature limits their applications. Some of the advanced catheter designs with active tip movement and controllable functions have been proposed and developed. Conducting polymers (CP) are also among the leading

designs active materials as they possess attractive features like large strain, low operating voltage and suitability for miniaturization [Wang *et al.*, (2007); Fang *et al.*, (2007); Lee *et al.*, (2009); John *et al.*, (2010)]. In contrast to the shape memory alloys, conducting polymers do not require high operating current [Haga *et al.*, (1998); Madden *et al.*, (2004)] making them particularly suitable for in-vivo biomedical applications. Furthermore, conducting polymers offer higher stiffness than IPMCs which is an important attribute often in catheter design [Carey *et al.*, (2004)].

Operator's skill is indispensable to insert and guiding the conventional catheter into a narrow and complex blood vessel or into the mouth for inspection and imaging, because the catheter and the guide wire do not possess the active actuation capability. However, for advanced navigation, active self actuation is needed for the catheter or the guide wire to move in the desired direction [Boppart *et al.*, (1997); Pan *et al.*, (2001)]. Further redundancy may be achieved by adding bending motion to the guide wire also. Thus, active guide wire system with self bending motion can navigate the conventional non-active catheters into a narrow and complex blood vessel as shown in the Figure 6.1. Figure 6.2 shows the photograph of the concept of the active catheter system. Forward and reverse mechanism makes the optical fiber along with the IPMC to move forward and backward so that the active length of the active catheter system changes. Further, employing a twisting control mode from outside of the body, the guide wire can move in multiple-directions even if it has only a one-directional bending functions. Current active catheter designs lack effective two dimensional control of motion. Such motions are important in procedures such as angiography, stent deployment, and aneurysms.

Electroactive polymers (EAPs) exhibit properties most closely matching those of natural muscles like low density, short response time, resilience and large actuation strains. As a result EAPs are sometimes called as artificial muscles. Among various types of EAPs, ionic polymer metal composites (IPMCs) are especially more prominent material to be used as the active catheter system because of their large deformation in the presence of very low voltage (1-2 V). IPMC actuators have also been suggested to be used for the active catheter application [Wang *et al.*, (2005)].

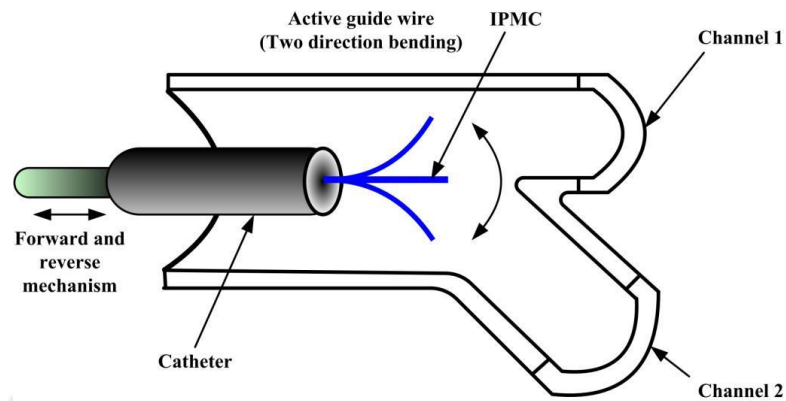


Figure 6.1 Mechanism of active guide wire system.

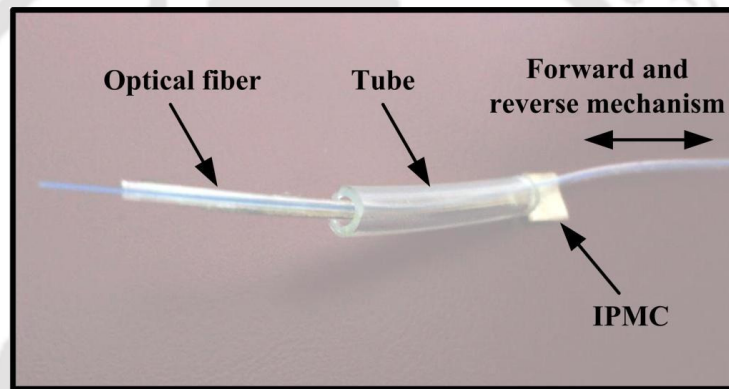


Figure 6.2 Photograph of the active guide wire.

In the present study, Ag-IPMC actuators have been chosen for their attractive properties such as, low actuation voltage, small size, high strain, compliance, biocompatibility, and ease of fabrication suitable for active catheter applications. A novel optical catheter probe with ionic polymer metal composite (IPMC) is developed for medical applications i.e., oral or inside the body imaging. An IPMC in a cantilever configuration is used to generate a transverse scanning movement with the application of low voltage (maximum 1.2V). Low driving voltage ensures safety for clinical application inside the body parts. The catheter design takes into account of the forward movement scanning mechanism make it suitable to take the scanning image covering more area at the same input voltage. The low cost Ag-IPMC in a cantilever configuration has potential applications in endoscopes for imaging of internal human tissues, blood vessel etc.

6.2 Preliminary design of a Forward Moving Active Catheter System

Optical coherence tomography (OCT) imaging technique nowadays is utilized for high resolution cross sectional imaging of highly scattered medium such as biological tissue [Huang *et al.*, (1991)]. OCT has been successfully used over the past few years for imaging of sectarian structures such as skin, retina, blood vessels, oral cavities, gastrointestinal tracts etc. This works on by linear scanning by an optical beam actuated across a target to create a 2D image. In the past, several actuation mechanisms are developed for actuation of optical fiber such as piezoelectric material in cantilever configuration [Kaneko *et al.*, (1995); Haga *et al.*, (2000)]. The use of thermoelectric actuator and an electrostatic actuator to swing a micro-electromechanical mirror and the rotation of electromagnetic devices (e.g., a galvanometer shaft by Bouma and Tearney (1999) and a micro-electromechanical motor by Tran *et al.*, (2004) were also reported. Eventhough, these actuation mechanisms have met the linear scanning requirement however, they typically require a relatively high driving voltage [Kaneko *et al.*, (1995); Lim and Lee (1999); Mineta *et al.*, (2001); Haga *et al.*, (2000)] or a complicated instrumental structure. Medical applications of OCT systems in endoscopic and intravascular imaging, low voltage is preferred for safety. In addition, the actuator must be simple in structure, biocompatible, non-acidic, rugged in handling, and suitable to work well within an endoscope.

An optical fiber is attached to the IPMC along the surface in such a way that, the bending actuation of the IPMC actuator generates the scanning movement of the optical beam. Optical fibers are very fine cylindrical glass fibers which allow light signals to travel from one end to the other end. The characteristics of fiber i.e., very light in weight, very flexible (easily twistable) makes it a natural choice in the medical field where bright lights needs to be focused on a target well within the body without a clear line-of-sight path. The catheter design takes into account of the forward movement scanning mechanism and thus it is possible to take scanning image covering wide range of area at the same input voltage. The schematic diagram of the active catheter with active and inactive portions is shown in Figure 6.3. The

lengths of the catheter which are inside of the tube are assumed to be the inactive. This portion of the catheter remains inactive during scanning operation. Figure 6.3 demonstrates how the length of the catheter is changing with the 1st increment.

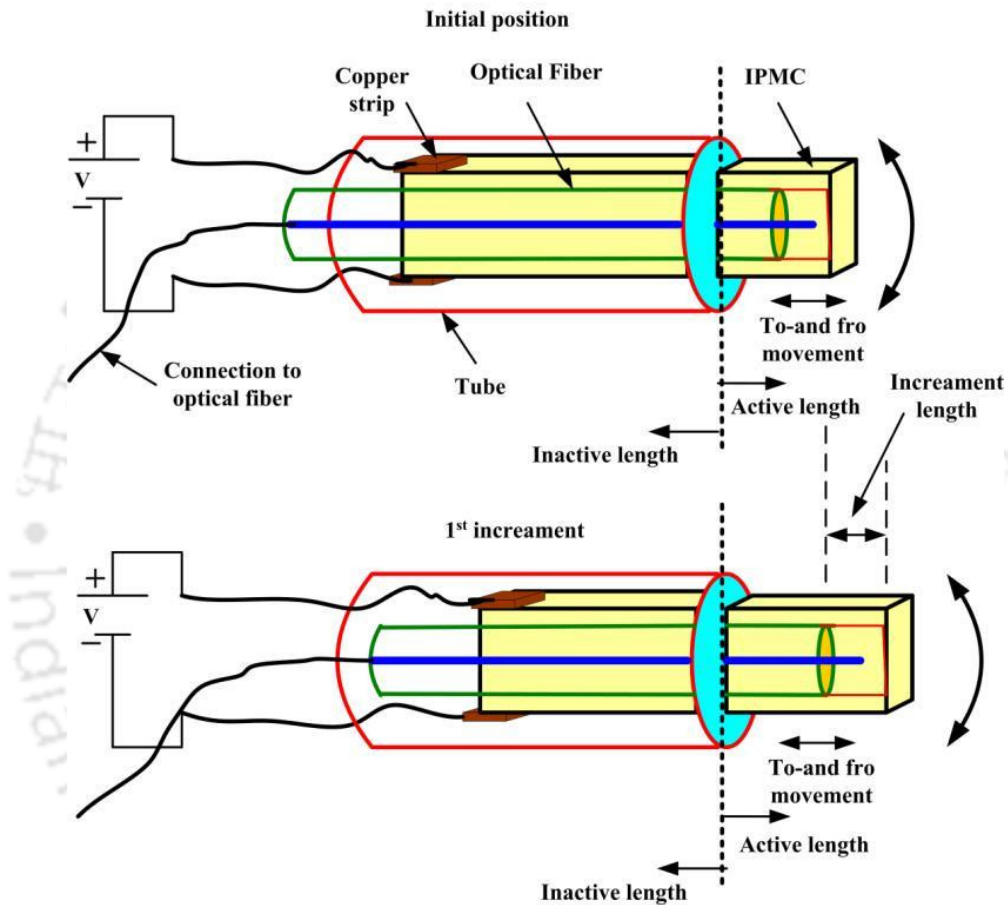


Figure 6.3 Schematics of preliminary designs of a forward moving active catheter with an optical fiber.

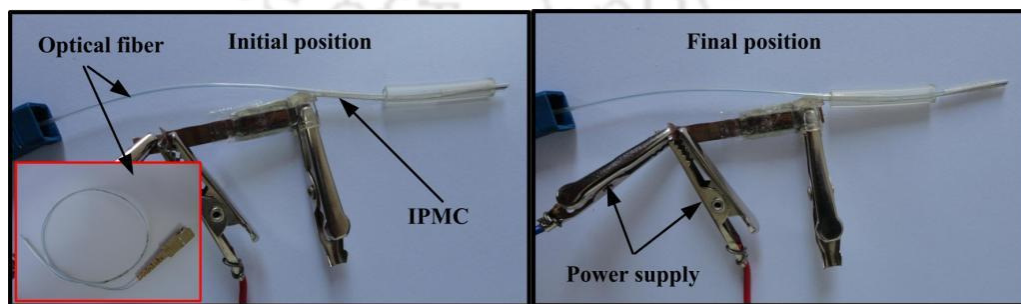


Figure 6.4 Photograph of the active catheter system.

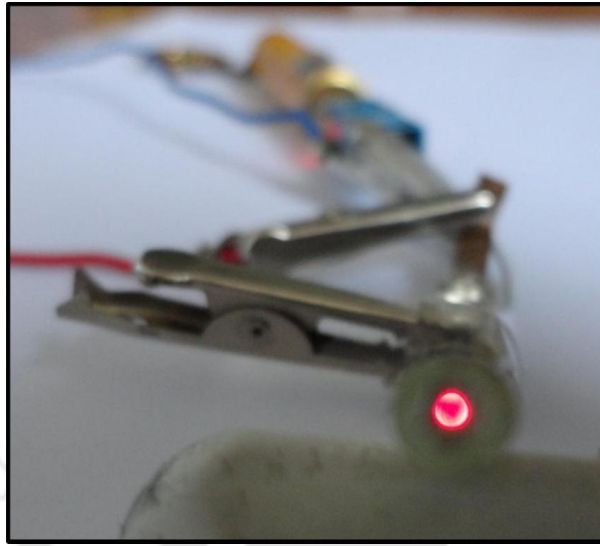


Figure 6.5 Photograph of the active catheter system with optical fiber.

The IPMC actuator with silver as the surface electrode is fabricated using chemical decomposition technique. The 0.2 mm thick IPMC membrane is cut into a 45 mm x 2 mm strip for active catheter system. Photograph of the catheter system with optical fiber is shown in Figures 6.4 and 6.5. A polytetrafluoroethylene (PTFE) hollow tube of small diameter (1 mm) is fixed on the tip of the IPMC strip by using glue. This PTFE tube is used to cover the tip of the optical fiber. Subsequently IPMC strip with optical fiber is connected with the power source. The whole arrangement is then inserted into a catheter tube. When an input voltage is applied, the bending motion of the IPMC actuator generates bending movement in the optical fiber and in a process the focusing light scan the affected area. The to-and-fro motion mechanisms of setup bestow the capability to scan more area. Thus the active mechanism is suitable for imaging the part of human oral cavity and also inside the body parts.

6.3 Mathematical Modeling

Photograph of the developed active catheter system with an optical fiber in working condition is shown in Figure 6.6. A schematic diagram of the IPMC actuator geometry in the catheter system is shown in Figure 6.7 while Figure 6.8 shows the catheter system inside the human body. The IPMC actuator has a uniform

rectangular cross section and it is fixed at one end. Euler-Bernoulli beam theory is assumed for modeling the free tip deflection. In order to describe the behavior of an IPMC actuator in a viscous/fluid medium such as water, the bending model of the active beam needs to explicitly take into account the interaction forces acting between the actuator and liquid medium.

Inside the fluid medium the actuator experiences a counter actuation forces due to the viscosity and due to its oscillation in the fluid. The effect of viscous/drag force is often neglect in air, but becomes significant when the actuator is operated in a denser medium such as water. As the actuator moves into the fluid medium, it also experiences a bouncy force as well. Here, both the forces i.e., drag and bouncy forces are taken into consideration in the formulation of bending moment generated in the active length of the catheter.



Figure 6.6 Photograph of the active guide wire with optical fiber in bending condition.

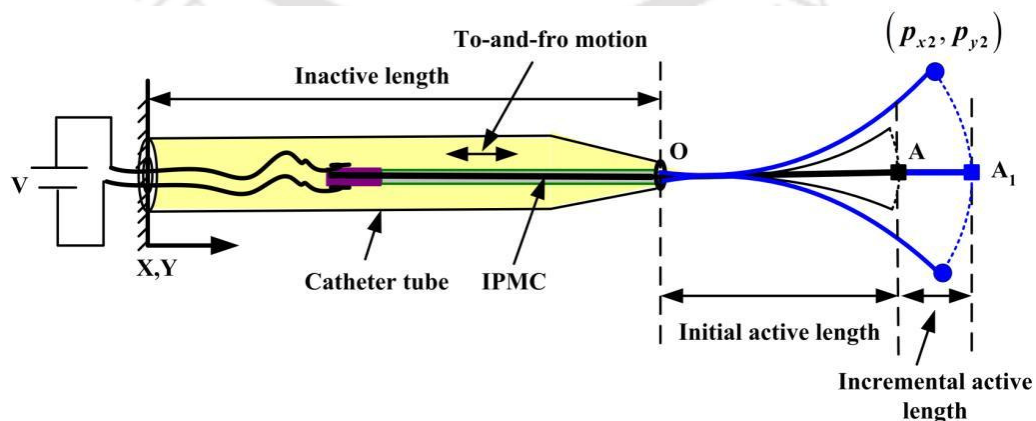


Figure 6.7 Schematic diagram of the active catheter system.

The drag force (F_{dg}) experienced by the actuator can be expressed as:

$$F_{dg} = \frac{1}{2} \rho_f v^2 C_D A \quad (6.1)$$

Where, ρ_f is the mass density of the fluid, v is the velocity of the object relative to fluid, A is the reference area, and C_D is the drag coefficient.

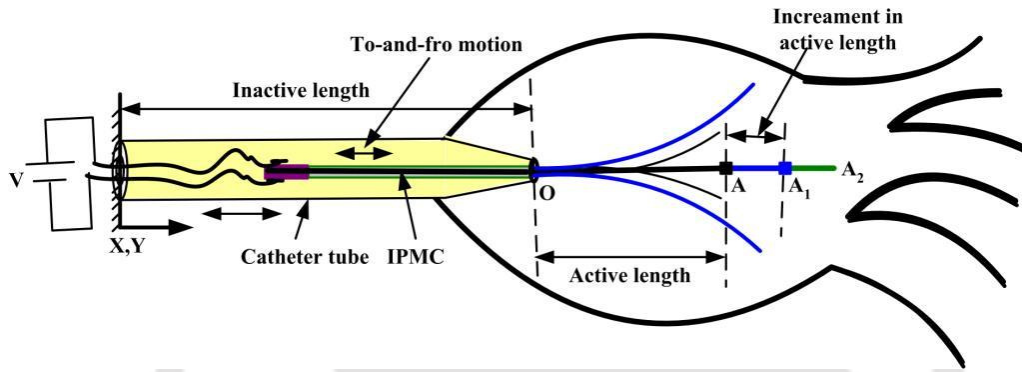


Figure 6.8 Schematic diagram of the active catheter system inside the body.

The bouncy force (F_b) developed is expressed as:

$$F_b = \rho_f V_{dis} \quad (6.2)$$

where, V_{dis} is the volume of the displaced fluid. Displaced volume of fluid can be expressed as: $V_{dis} = l_a \times w_d \times h$, where l_a is the active length, w_d and h are the width and thickness of the IPMC actuator. Thus, the effective bending moment generated by the IPMC actuator in a fluid medium (M_e), can be expressed as:

$$\begin{aligned} M_e = & M_0 [H(X - x_i) - H(X - x_{i+1})] \\ & - \left\{ \left(\frac{1}{2} \rho_f v^2 C_D A \right) (H(X - x_i) - H(X - x_{i+1})) \right\} \\ & - \rho_f l_a w_d h \{ H(X - x_i) - H(X - x_{i+1}) \} \end{aligned} \quad (6.3)$$

Where, M_0 is the bending moment generated in air, X and x_i denote the length of the actuator along the link length. $i = 1, 2, 3, \dots, n$ indicates number of increments.

$[H(X - x_i) - H(X - x_{i+1})]$ is the Heaviside function and is defined as:

$$[H(X - x_i) - H(X - x_{i+1})] = 1 \quad \text{for } la > 0$$

$$[H(X - x_i) - H(X - x_{i+1})] = 0 \quad \text{for } la \leq 0$$

Subsequently, tip position $A(p_{x1}, p_{y1})$ of the actuator is obtained as:

$$p_y = \frac{EI}{M_e}(1 - \cos \varphi) \quad (6.4)$$

$$p_x = \frac{EI}{M_e} \sin \varphi \quad (6.5)$$

where, EI is the flexural rigidity and φ is the tip angle of IPMC.

6.4 Experiments

To investigate and study the working of catheter mechanism in fluid medium, experiment is conducted in underwater condition. At first, an IPMC actuator of size $20 \times 5 \times 0.2$ (mm³) is experimented in underwater by applying an input of 1.0V as shown in Figure 6.9. Figure 6.9(a) shows the static condition before the voltage is applied and Figure 6.9(b) shows when an input of 1.0V is applied. It is observed that the distance travelled by the IPMC actuator within water is about 9.0 mm while in air it is 9.4 mm. Active catheter system is assembled using the optical fiber which is glued over the surface of the IPMC actuator. The whole assembled catheter system is then placed within the water. Figure 6.10 demonstrates the active catheter system in underwater environment. The voltage is provided on the other end of the IPMC actuator through a copper strip which is clamped by a clip.

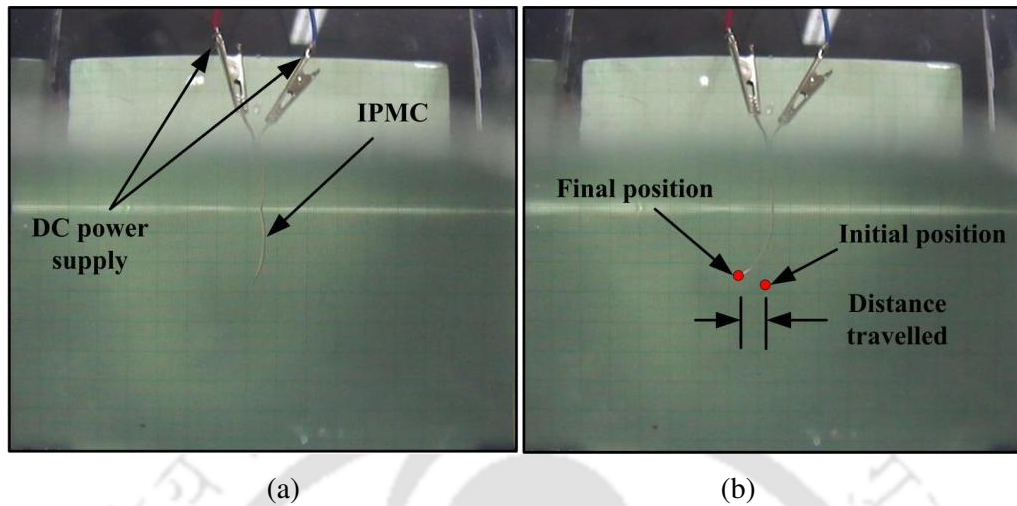


Figure 6.9 Actuation of the Ag-IPMC actuator (a) initial condition (b) input of 1.0V DC in water.

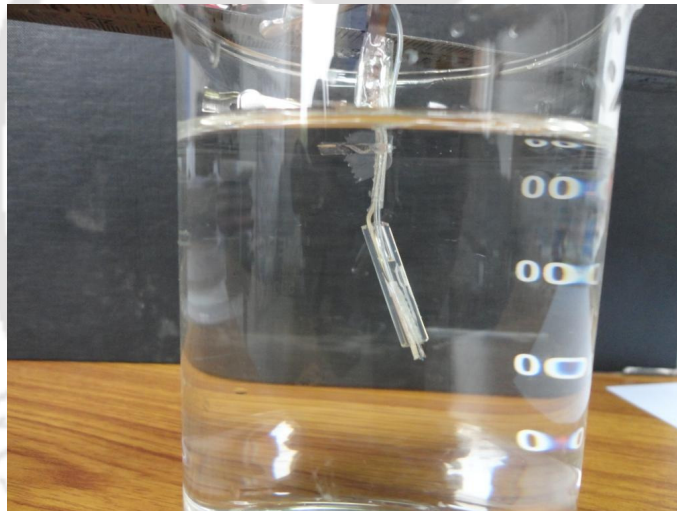


Figure 6.10 Photograph of the active catheter system in underwater.

6.5 Simulation Results

Simulations are done for a single patch IPMC actuator taking into account of the data as given in Table 3.1. In the simulation, three steps results are shown for the active length of the actuator while initial active length is taken 20 mm. Active length is then increased by 5 mm for each step. Figure 6.11 and 6.12 show the simulation results in open air and under water, respectively.

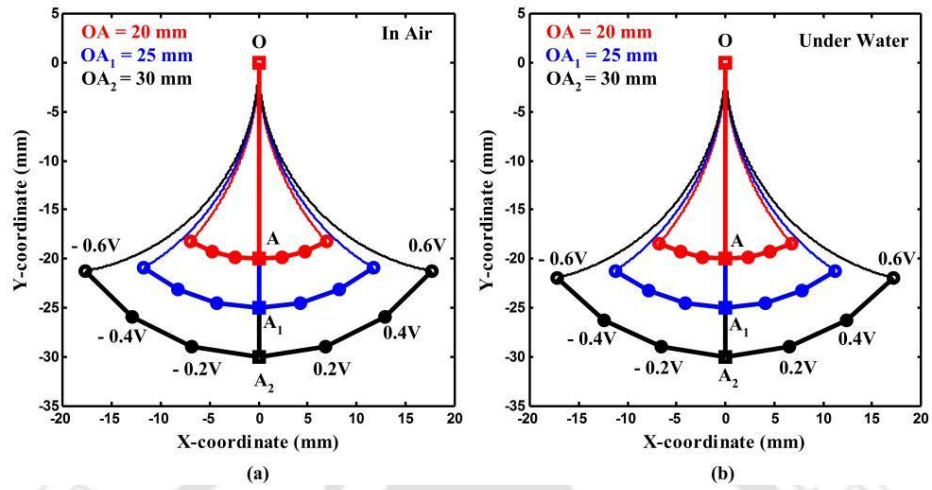


Figure 6.11 Total path traversed by the IPMC actuator for an input of ± 0.6 V (a) in open environment (air) (b) underwater.

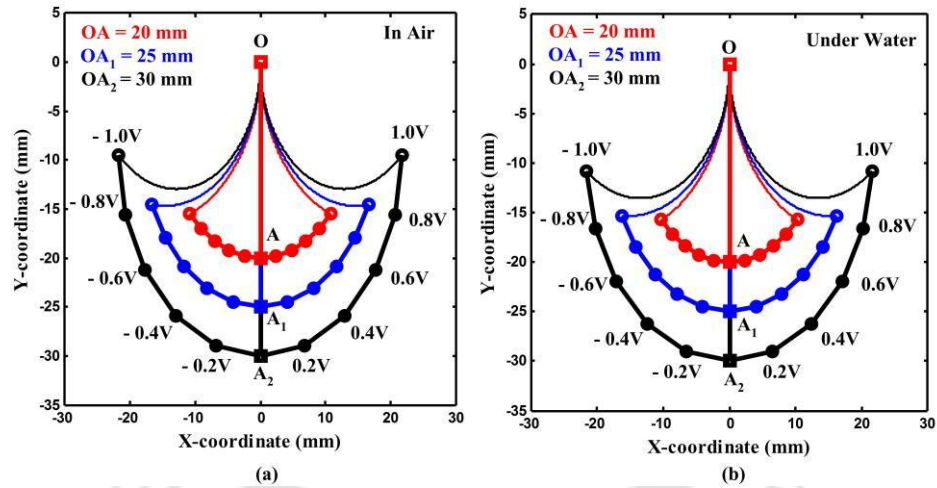


Figure 6.12 Total path traversed by the IPMC actuator for an input of ± 1.0 V (a) in open environment (air) (b) underwater.

Table 6.1: Tip position of the IPMC actuator in open environment.

IPMC active length	Tip position	Applied voltage					
		0.2V	0.4V	0.6V	0.8V	1.0V	1.2V
Initial (20 mm)	p_x	2.4211	4.7709	6.9809	8.9880	10.7372	12.1832
	p_y	19.8033	19.2200	18.2707	16.9889	15.4192	13.6158
1 st increment (25 mm)	p_x	4.2176	8.1928	11.7018	14.5562	16.6173	17.8054
	p_y	24.5192	23.1099	20.8691	17.9495	14.5470	10.8852
2 nd increment (30 mm)	p_x	6.8274	12.9348	17.7034	20.7009	21.7375	20.8836
	p_y	28.9378	25.8861	21.2289	15.5422	9.5077	3.8136

Table 6.2: Tip position of the IPMC actuator in underwater environment.

IPMC active length	Tip position	Applied voltage					
		0.2V	0.4V	0.6V	0.8V	1.0V	1.2V
Initial (20 mm)	p_x	2.3270	4.5908	6.7302	8.6889	10.4167	11.8721
	p_y	19.8183	19.2793	18.4005	17.2103	15.7473	14.0581
1 st increment (25 mm)	p_x	4.0358	7.8593	11.2734	14.1091	16.2378	17.5791
	p_y	24.5603	23.2688	21.2069	18.5033	15.3243	11.8621
2 nd increment (30 mm)	p_x	6.5342	12.4383	17.1629	20.3094	21.6770	21.2824
	p_y	29.0292	26.2294	21.9232	16.5985	10.8423	5.2620

6.5.1 Calculation of Scanning Path

The path traversed by the active catheter gives measure of the scanning area. Thus finding out the path traversed by IPMC actuator for each step is important for

comparative study. The incremental length of the catheter, results in further increase in scanning path, thus scanning area. Figure 6.13 shows the geometric configuration of the path traversed by IPMC. OA is the initial length of the actuator before activation. A' and A'' are the tip position (p_x, p_y) of the actuator and l_{ei} is the effective length of the curvature for an input voltage V. θ_1 and θ_2 are the angles between the initial position and after application of input voltage V and l is the length of the actuator. S_i is the path travelled by the tip of the actuator. The effective length of the curvature (l_{ei}) can be calculated as:

$$l_{ei} = \sqrt{(p_{xi})^2 + (p_{yi})^2}, \quad i = 0.2, 0.4, 0.6, 0.8, 1.0, 1.2V \quad (6.4)$$

The angle between the initial position and after application of input voltage (θ) can be calculated as:

$$\theta_i = \tan^{-1} \left(\frac{p_{xi}}{p_{yi}} \right) \quad (6.5)$$

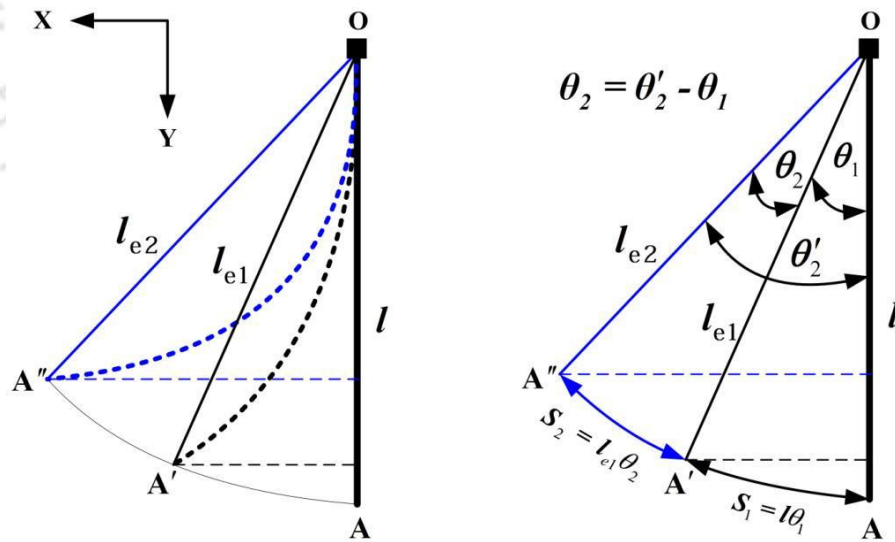


Figure 6.13 Diagram for scanning path calculation.

Table 6.3 Scanning path calculated for various input voltage at initial condition (active length=20 mm).

Voltage (V)	In air				Under water			
	l_{ei} (mm)	θ_i (rad)	S_i (mm)	Total path ($2S_i$)	l_{ei} (mm)	θ_i (rad)	S_i (mm)	Total path ($2S_i$)
0.2	19.9507	0.1217	2.434	4.868	19.9544	0.1168	2.336	4.672
0.4	19.8032	0.1216	2.426	4.852	19.8183	0.1169	2.332	4.664
0.6	19.5589	0.1217	2.410	4.82	19.5927	0.1169	2.316	4.632
0.8	19.2199	0.1216	2.378	4.756	19.2793	0.1169	2.290	4.58
1.0	18.7893	0.1217	2.339	4.678	18.8808	0.1169	2.253	4.506
1.2	18.2707	0.1216	2.284	4.568	18.4004	0.1168	2.205	4.41

Thus, for an active length of 20 mm, the actuator travelled 28.5 mm for $\pm 1.2V$ in open environment (in air) and 27.464 mm in under water conditions.

Table 6.4 Scanning path calculated for various input voltage at initial condition (active length=25 mm).

Voltage (V)	In air				Under water			
	l_{ei} (mm)	θ_i (rad)	S_i (mm)	Total path ($2S_i$)	l_{ei} (mm)	θ_i (rad)	S_i (mm)	Total path ($2S_i$)
0.2	24.8792	0.1703	4.257	8.514	24.8896	0.1628	4.07	8.14
0.4	24.5191	0.1703	4.236	8.472	24.5602	0.1629	4.054	8.108
0.6	23.9259	0.1704	4.178	8.356	24.0171	0.1628	3.998	7.996
0.8	23.1099	0.1703	4.074	8.148	23.2688	0.1629	3.912	7.824
1.0	22.0850	0.1704	3.937	7.874	22.3271	0.1629	3.790	7.58
1.2	20.8691	0.1703	3.761	7.522	21.2069	0.1628	3.634	7.268

It is observed that with increment of active length i.e., from 20 mm to 25 mm, the scanning path has increased from 28.5 mm to 48.8 mm in open air while from 27.4 mm to 46.9 mm in under water for the same input voltage $\pm 1.2V$. Figure 6.14 shows the scanning path length for single sequence of input voltage for various active lengths of the actuator. It is observed that as the active length increases the

scanning path also increases with the same input voltage. Figure 6.15 shows the total scanning path with various active lengths.

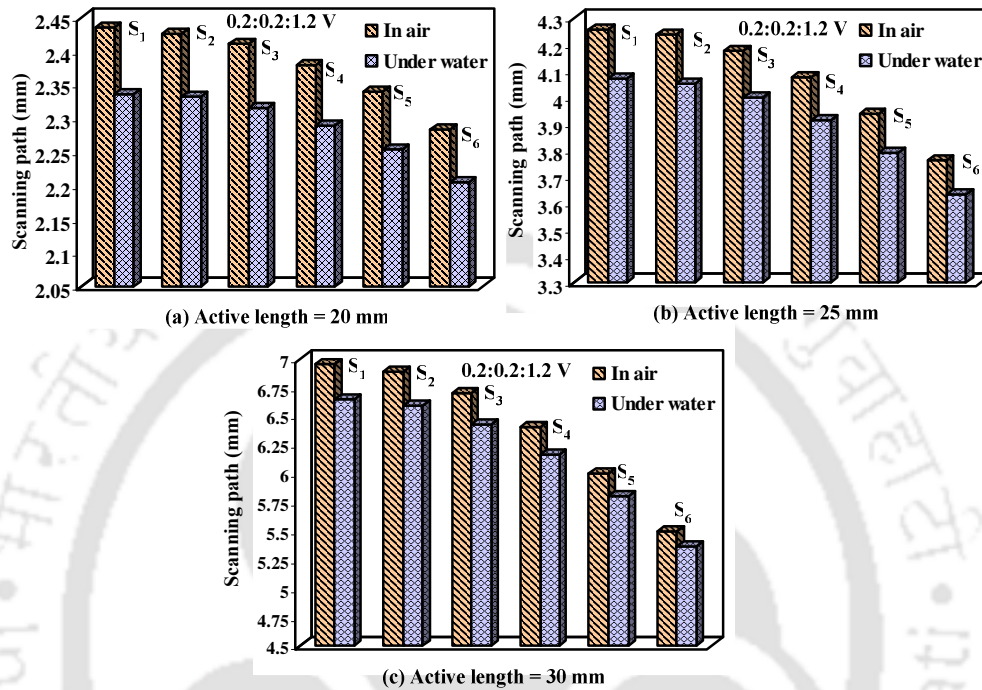


Figure 6.14 Scanning paths for single sequence of input voltage for various active lengths.

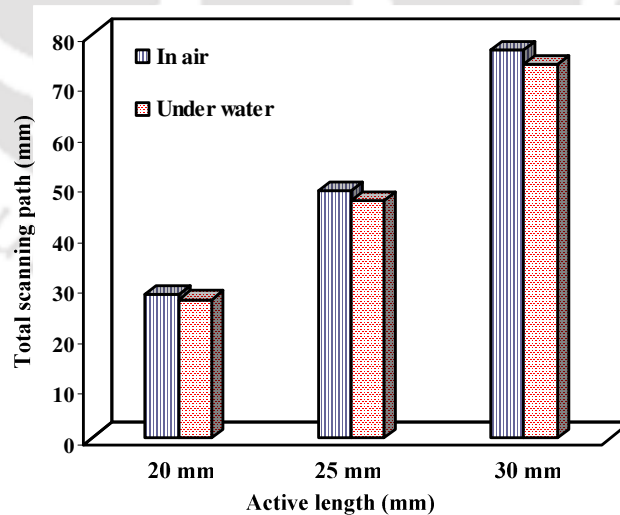


Figure 6.15 Comparison between total scanning paths for different active lengths for both positive and negative sequence of input voltage.

Table 6.5 enlists the incremental scanning path due to increase of active length. It is clearly observed that scanning path significantly increased due to increase of active length for the same input voltage.

Table 6.5 Incremental scanning path in open air for various input voltages.

Voltage	Incremental scanning path		
	Initial ($l_e = 20\text{ mm}$)	1 st increment ($l_e = 25\text{ mm}$)	2 nd increment ($l_e = 30\text{ mm}$)
$\pm 0.2\text{V}$	0	3.64	8.41
$\pm 0.4\text{V}$	0	7.26	17.95
$\pm 0.6\text{V}$	0	10.8	26.54
$\pm 0.8\text{V}$	0	14.19	34.59
$\pm 1.0\text{V}$	0	17.38	41.9
$\pm 1.2\text{V}$	0	20.34	48.33

6.7 Summary

An active variable length catheter system has been developed and analyzed. Mathematical modelling of the mechanism has been done taking into account of the effect of viscous damping and drag force. Simulation has been performed for open air and under water conditions. Experiments are conducted with an IPMC actuator under water in variable active length mode to understand the resistance encountered during actuation and the results are verified with the simulation results. The novelty of the active catheter mechanism is that the length of the IPMC actuator varies with input voltage thus can cover wide range of working area with same input voltage. Apart from oral and dental cavity inspection, the low cost Ag-IPMC mechanism has the potential applications in endoscopes for imaging of internal human tissues, blood vessel etc.

Conclusions and Scope for Future Work

7.1 Conclusions

In the present work, a low cost ionic polymer metal composite (IPMC) actuator of silver (Ag) electrode has been fabricated. Several steps of trial-and-error attempts results in successful fabrication of the actuator following the silver mirror reaction (chemical decomposition) process. Several experiments have been carried out to characterize the developed IPMC actuator. The effectiveness and performance of the actuator is studied and compared with the noble metal such as gold, platinum based IPMCs. The experimental investigation and comparative study demonstrates superior actuation capability of the actuator. Multiple experiments are conducted to ascertain the effect of dehydration on actuation capability of the IPMCs. An empirical model has been developed subsequently which estimates the continuous dehydration of IPMC actuator during working condition.

Further, extensive bending and vibration tests have been performed under a wide range of DC and AC input voltage. The experimentally obtained tip position data are used to develop theoretical models for prediction of linear and nonlinear vibration response of IPMC actuator under both DC and AC voltage actuations. These responses are also obtained by incorporating loss due to dehydration. The developed general purpose theoretical models can be used for any IPMCs to predict the continuous moisture loss during actuation and to study the effect of dehydration on vibration response. Finally, an active catheter system has been developed to demonstrate the applicability of the developed IPMC in biomedical application.

The specific conclusions of the present work can be summarized as follows:

- Ionic Polymer Metal Composites (IPMCs) with silver (Ag) as surface electrode is successfully fabricated following silver mirror reaction (chemical decomposition) method. Morphological characteristics of the developed IPMC examined by scanning electron microscope illustrates that there is good bonding between the metal and polymer layer and a uniform densely packed distribution of Ag particles is found over the polymer surface. Also, the developed IPMC shows no oxidation problem during actuation in open environment. The coefficient of thermal expansion (CTE) of IPMC shows negative value with low temperature, and changes from negative to positive value with rise in temperature. DSC and TGA tests of the IPMC confirm improved thermal stability, and micro-tensile test confirms the increase in Young's modulus. A typical operational life of the developed IPMC of dimension $20 \times 5 \times 0.2$ (mm³) is around 90-100s in open air environment subjected to 0.6V DC. The actuator can work again upon full hydration by keeping it into deionized water. The cost of fabricating a $40 \times 40 \times 0.2$ (mm³) Ag-IPMC is found to be around Rs.2500.00 (50 USD) which is much less compared to the conventional gold, platinum based IPMCs. The developed IPMC shows superior performance such as large bending deflection, less surface resistance, larger tip force compared to other electrode materials. The experimental investigation, characterization and performance of the developed IPMC demonstrate its effectiveness to be used as actuator and artificial muscle materials.
- A new term 'loss-factor' is introduced in this work to estimate the loss of moisture from IPMC during working condition. A simple but robust empirical model has been developed from the experimental data to predict the loss of water content from IPMC actuator in terms of input voltage and time in the open working environment. The proposed model is robust and valid for any IPMC actuator in working medium that experiences continuous dehydration. The estimation helps to understand the amount of input voltage or the moisture quantity needs to be added with time to compensate the loss. Forward bending

model is developed for segmented IPMC patches following both the fixed and variable parameters pseudo-rigid body modelling technique and the dehydration effect is assessed on tip positioning and bending resistance. The proposed model is helpful for obtaining the tip position by compensating the loss due to dehydration.

- Both theoretical and experimental investigations confirm that both bending tip deflection and amplitude of vibration response decreases with dehydration. The numerically obtained steady-state response is found to be in good agreement with the experimental results. The proposed dehydration model and experimental results are useful for obtaining the tip position accurately by attenuating the vibration and compensating the loss due to dehydration.
- A closed form expression for the response amplitude and detuning parameter has been developed which can be used to obtain the nonlinear frequency response curve for any IPMC actuator. The simplified expression can be used for finding the response of the system instead of solving the temporal equation of motion which is time consuming and requires more memory space. The numerical results are compared and verified by performing experiments and the results are found to be in good agreement. It is concluded that in micro-scale actuator, amplitude of vibration could increase considerably when a small change in excitation amplitude or input voltage occurs. Further, in micro-scale applications, the amplitude of vibration can be easily increased and it may bring the system into non-linear regime, and therefore, it is useful to utilize the non-linear analysis to predict the behavior of the system more accurately.
- The novelty of the active catheter mechanism is that the length of the IPMC actuator varies with input voltage thus can cover wide range of working area with same input voltage. Apart from oral and dental cavity inspection, the low cost Ag-IPMC mechanism has the potential applications in endoscopes for imaging of internal human tissues, blood vessel etc.

7.2 Scope for Future Work

The research work reported in this thesis represents significant advancements in the field of fabrication, characterization, modeling of IPMC actuator. Although, IPMC applications in various fields and several other actuator related issues demand further investigation. Some scopes of further work are listed below.

- The fabrication procedure can be further optimized by understanding the physical actuation mechanism by exploring different ionic liquids as solvents, different ionomer as the base polymer and varying the electrode materials. Further, fabrication procedures such as PVD or hot press techniques may be employed to fabricate thicker and multilayered IPMC. Similar analyses carried out in this work may be used for analyses of multilayered IPMC and to study their effects on the actuation performance of Ag-IPMC.
- A similar study can be carried out to investigate the sensing properties for Ag-IPMC actuator. This thesis focused mainly on the actuation of the IPMC transducer; while sensing is also an important application area of IPMC transducers.
- In this work nonlinear vibration analysis has been carried out considering single frequency excitation. One may consider multi-frequency excitation to analyze the system.
- The concept and development of the active catheter system is at the primary stage needs further working and improvement for successful implementation. Controlling of active catheter system can further be explored along with other biomedical applications of IPMCs.
- Analysis and compensation for delay to the system response.

Publications from the Present Thesis

Journals

1. Dillip Kumar Biswal, Dibakar Bandopadhyaya, and Santosha Kumar Dwivedy, (2010), Evaluation of dehydration loss and investigation of its effect on bending response of segmented IPMC actuators, *International Journal of Smart and Nano Materials*, Vol.1, issue 3, pp. 187-200.
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