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Technical note

Electroactive polymeric sensors in hand prostheses: Bending response of an ionic polymer metal composite

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Abstract

In stark contrast to the inspiring functionality of the natural hand, limitations of current upper limb prostheses stemming from marginal feedback control, challenges of mechanical design, and lack of sensory capacity, are well-established. This paper provides a critical review of current sensory systems and the potential of a selection of electroactive polymers for sensory applications in hand prostheses. Candidate electroactive polymers are reviewed in terms of their relevant advantages and disadvantages, together with their current implementation in related applications. Empirical analysis of one of the most novel electroactive polymers, ionic polymer metal composites (IPMC), was conducted to demonstrate its potential for prosthetic applications. With linear responses within the operating range typical of hand prostheses, bending angles, and bending rates were accurately measured with 4.4 ± 2.5 and $4.8 \pm 3.5\%$ error, respectively, using the IPMC sensors. With these comparable error rates to traditional resistive bend sensors and a wide range of sensitivities and responses, electroactive polymers offer a promising alternative to more traditional sensory approaches. Their potential role in prosthetics is further heightened by their flexible and formable structure, and their ability to act as both sensors and actuators.

Keywords: Electroactive polymers; Ionic polymer metal composites; Prosthetics; Bend sensor; Sensory; Upper limb prostheses; Hand; Biomimetic

1. Introduction

In a world of unknown, the hand is a key avenue for exploration and manipulation. From our first attempts at tool-making to the performance of Beethoven's concertos, its development has hastened us along the evolutionary pathway of success [1]. Key to the hand's functionality and dexterity is its complex sensory system, which provides the requisite information to bridge our inner neural networks to our ever-changing environment. With receptors honed to pressure, vibration, temperature, and spatial positioning, this complex system enables us to catch the glass vase as it slips from grasp, to drop the hot potato before damage is done and to engage in a firm handshake at the office and yet hold a child's hand at home.

An entourage of technologies is positioned to benefit from the realization of engineered sensory systems with capabilities emulating that of the natural hand, ranging from remote manipulation in research or space exploration [2], to non-invasive surgical techniques [3], to creation of virtual environments for training simulations [4–6]. Perhaps first in line, however, is the field of upper limb prosthetics in which sensory feedback is critical for obtaining adaptive grasp, reflex capabilities, prevention of slip and tactile exploration [7–16]. Currently, functional limitations and challenges of unreliable control have motivated over 70% of amputees in the United States to select hooks for value of their functionality [17] while 30–50% do not use their hand prostheses on a regular basis [18]. One key limitation responsible for the marginal performance of prostheses in use today is the lack of sensory feedback available to the controller and the user, hindering the ability to respond in a natural and appropriate manner in accordance with the external environment [7,8,19–22].

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The purpose of this study is to review the current literature pertaining to sensory systems in prosthetic hands, to characterize physical and biomimetic sensory properties of a range of electroactive polymers (EAPs), and to empirically investigate the potential of one candidate EAP for sensing bending angles and rates in the context of a prosthetic hand.

2. Sensory system in prosthetic hands

In contrast to the wide ranging sensitivity of the natural hand, prosthetic hands are typically limited to a single force sensor on the thumb and a strain sensor incorporated into the transmission system to regulate grasp and to protect the drive mechanism from damage [23,24]. At present, there is no clinically acceptable system available to provide a sense of touch in the prosthetic hand [7]. Sensory feedback is therefore largely limited to that of a visual nature, together with more subtle auditory cues such as the sound of speed changes of the motor [25]. Evidence indicating the importance of sensory feedback is widespread with haptic feedback reportedly enhancing functional performance by 50% in virtual environments [6]. Sensory feedback is fundamental to adaptive grasp and slip prevention as necessary for tasks, such as precision grips, which encompass an estimated 60% of activities of daily life [26]. Although information from the visual system aids in adaptive grasp both in the natural and prosthetic hand [27], grip forces in the absence of direct sensory feedback tend to be much higher than necessary in activities of daily life such as eating [1,19]. In prosthetics, where limiting power consumption and effort are of significant importance, this becomes a considerable drawback.

Recent efforts have acknowledged the importance of sensory systems and feedback to the functionality of a prosthetic hand leading to various advancements in this field. For instance, the commercially available Otto Bock SensorHand provides an AutoGrasp feature intended to prevent object slippage by regulating grip using feedback from force sensors sensitive to shifts in the centre of gravity of grasped objects [28]. The Cyberhand [24], currently in development, is perhaps one of the most elaborate sensory endeavours with Hall effect position sensors to determine joint angles, tension sensors on the flexion activating cables for force measurements, and an accelerometer inside the palm to determine contact with handheld objects. In addition, flexible on–off touch sensors to determine contact points, and three-dimensional force sensors are distributed throughout the hand and fingertips to provide tactile sensation. Another popular approach is the use of a "sensory glove" such as that developed by Mingrino et al. [9] which incorporates vibration and force sensors embedded in an elastomer glove. Along similar lines, is the development of a sensor thumb [21] which employs a matrix of force sensing elements on the thumb to provide information for closed loop control during a variety of grasp configurations typical of tasks of daily living. The devices designed in each of these developmental approaches tend to be quite bulky and aesthetically unnatural, a recurring challenge beleaguering artificial sensors. Recent developments have however realized sensory detection on a scale typical of a fingertip with thick-film force sensors for static measurements up to 100 N, screen-printed piezoelectric vibration sensors for slip detection, and temperature sensors to warn of potentially damaging extremes, and provide temperature compensation for the sensory components [29].

Several additional studies are also in progress as relevant to the provision of sensory feedback in prosthetic designs and are referenced for interested readers [2,8,12,13,15,16,30–33]. Typically, the most commonly used sensors include potentiometers, accelerometers, Hall effect sensors, strain gauges, and piezoelectric vibration sensors. Progress in this area has been limited firstly by a lack of appropriate sensors and means of affixing them to commercially available prosthetic hands and secondly, by the ability to set-up a multi-channel communication link for increased control [7].

3. Electroactive polymers as biomimetic sensors

With 17,000 tactile receptors [34] alone and the capacity for processing the immense quantities of sensory information generated thereby, the natural hand has proven a daunting model to emulate despite the plethora of sensors currently on the market. For a comprehensive review of sensors and sensor selection, the interested reader is referred to Shieh et al. [35] for a general review of traditional sensors and Webster [36] for those specific to biomedical applications. Many traditional approaches, however, are not suitable for implementation in a prosthetic hand, which demands a flexible, compact, and non-intrusive packaging harbouring a highly sensitive sensor with fast response and continuously variable output while simultaneously addressing issues of power consumption, cost, and durability [37].

An electroactive polymer (EAP) is a material that exhibits a measurable response to a given stimuli, most commonly a change in shape or voltage potential. A myriad of EAPs with varying sensitivities and responses exist as a result of a range of distinct chemical and thermodynamic structures. A comprehensive book by Bar Cohen on the subject matter provides an excellent compilation of information regarding electroactive polymers in general [38].

EAPs are lightweight, pliable, quiet, and shatterproof, boasting electromechanical properties that can be tailored to meet the needs of specific applications. In addition, electroactive polymers offer many of the typical advantages of polymers, such as ease of manufacture and formability. As such, EAPs have currently been implemented in a host of sensory applications ranging from haptic and neural interfaces and artificial noses [39] to intelligent chemical sensing systems [40] to measurement of blood pressure and pulse rates [41]. The following will detail a selected few electroactive

Table 1
EAP sensors

Sensor	Principle of operation	Advantages	Related applications
Polypyrrole	Oxido-reduction reactions induce change in material properties, i.e., volume, resistivity	Biocompatible Multiformable (films, fibres, tubules, sheets) Reacts to electrical, mechanical, thermal, chemical stimulation	Strain gauges and sensing fabrics Sensory gloves for: (a) monitoring of hand movement and positioning; (b) assistive movement of the fingers for rehabilitative purposes
PVDF	Piezoelectric effect induces electrical response to mechanical deformation and vice versa	Ease of manufacture Sensitive to vibrations	PVDF vibration and contact sensors in robotics and prosthetics applications for tactile discrimination or slip detection
IPMC	Shift of mobile charges induced by deformation, resulting in a charge imbalance	Sensitive to large deformation bends Exhibits biomimetic response Sensitive to moisture content, metal content and distribution	Tactile sensors Exploratory and new

polymers with high potential in terms of sensory provision for prosthetic hands and the current applications in which they have been implemented. Table 1 provides a summary of the most prevalent EAPs as biomimetic sensors.

3.1. Polypyrrole

A subset of the conductive polymer class, polypyrrole is an emerging candidate in the race towards practical artificial muscles and sensors. It is biocompatible, can be fashioned in the form of films, fibres, tubules, and sheets, and enacts a change in volume or electrical conductivity due to the transport of ions or solvent resulting from oxidation/reduction reactions, as typical of conductive polymers. It responds in the presence of an applied electrical field, mechanical deformation or even thermal or chemical stimuli. Material response characteristics are highly dependent on the chemical structure of the material.

Polypyrrole fibres have been incorporated into fabrics enabling conformable, wearable sensor mechanisms with high thermoresistive and piezoelectric coefficients [42,43]. These fabrics are particularly useful as strain gauges as their resistivity changes markedly when stretched and is dependent on stretch direction [43]. With a comparable gauge factor of 2 and an increased dynamic range ten times that of conventional strain gauges, these conducting fabrics have been fashioned into sensory gloves and have been demonstrated for the monitoring of hand movement and positioning [43,44]. Conversely, when used in actuation mode (i.e., an electrical stimuli is applied to obtain a physical expansion/contraction), these gloves have been employed for assistive movement of the fingers for rehabilitative purposes [42].

Based on this sensory application, a parallel could be drawn between these conducting fabrics and the functionality of the slowly adapting, type 2 afferents (SAII) of the natural hand. The SAII afferents contribute to proprioception (i.e., a sense of spatial positioning and velocity) through the detection of skin strains, which can be related to joint angles [45,46] and used to sense position and velocity as well as muscle lengths and generated forces [7]. Activities of daily living incur skin strain of 10–15% over the back of the hand [47]. Conductive fabrics consisting of polypyrrole fibres respond much in the same manner as the rapidly adapting (RA) afferents of the hand in terms of temporal adaptation and are not conducive for detecting slowly varying strains [48]. Conversely, conductive fabrics composed of carbon loaded rubber (CLR) as opposed to polypyrroles are suitable for the detection of steady and slowly varying strain levels (bandwidth of dc to 8 Hz) and adapt in a manner similar to the SA mechanoreceptors [44].

CLR also exhibits high potential for prosthetic applications with qualities, such as manufacturability, flexibility, small size, and resilience. Conversely, disadvantages of CLR sensors include hysteresis, creep, and instability. CLR has been used in a variety of force and pressure transducers for tactile applications, for contact detection, and for grasp control in both robotic and prosthetic hands [14,32,48,49].

3.2. Polyvinylidene fluoride

The strong piezoelectric characteristics of the polymer, polyvinylidene fluoride (PVDF), have been widely exploited for vibration and contact transduction in applications ranging from smart structural components to ultrasonics. Often in the form of films, PVDF is mechanically drawn to orient its molecules and polarized to yield strong piezoelectric behaviour. Thus, the application of an electric field across the thickness of the film (in the direction of polarization) will cause a decrease in thickness due to the reorientation of its molecules and net polarization. Conversely, application of a physical force will yield an electrical response at the surface of the polymer.

PVDF vibration and contact sensors have been implemented in a number of robotics and prosthetics applications, most commonly for tactile discrimination or slip detection [50–55]. Tada et al. [52] for example, fabricated a soft fingertip for tactile discrimination using PVDF-based sensors imbedded in different silicone rubber layers of the fingertip in an attempt to mimic the natural cutis and epidermis layers of the skin. The PVDF sensors responded to transient/low strain stimuli with the upper situated sensor exhibiting a strong vibratory response to the stimuli. The deeper situated PVDF sensors experienced a filtered version of the stimuli, as the silicon layer acted as a low pass filter, and thereby responded with a more stable signal. By processing the differences in the signal responses of the PVDF sensors at different layers, the ability to discriminate between wood and paper textures was demonstrated. Fujimoto et al. [54] further used PVDF in order to mimic the characteristics of the RA afferents in the skin. Using these sensors and developed neural networks, they designed a system to determine incipient slip. Despite challenges in calibrating sensors based on soft platforms, the advantages of form, ease and flexibility of fabrication, and dynamic sensory response are affording these piezoelectric polymers with increasing popularity.

3.3. Ionic polymeric metal composites

Ionic polymer metal composites (IPMC) hold much promise in the development of artificial muscles and "smart" sensors. Composed of a thin polymeric material sandwiched between two plated metal electrodes, the materials produce a voltage on the order of millivolts when mechanically deformed (i.e., in bending, stretching, and compression). It is believed that this voltage potential is generated by a shift of mobile charges caused by material stresses induced by deformation, resulting in a charge imbalance [56]. The performance of IPMCs depends on a number of factors including water content and swelling, metal content and distribution, together with molecular and physical properties of the polymer [57]. For details regarding the chemical structure, manufacturing, and modeling of these materials, the interested reader is referred to a series of review papers compiled by Shahinpoor and Kim [58–60].

Previous developments in IPMC sensors have demonstrated promising results in a variety of applications. Ferrara et al. [61] for example, implemented an IPMC pressure transducer for load measurements in the spine. Reduced thickness, biocompatibility, and the ability to operate in wet environments are ideal properties of IPMC sensors for biological environments in which traditional sensors are often lacking. Konyo et al. [62] proposed a tactile sensor based on four IPMC modules used for velocity and directional detection. A method of patterning an IPMC film to enable dual sensing and actuating capabilities was also suggested. A final example of IPMCs in action was presented by Keshavarzi et al. [63] who successfully employed these sensors for blood pressure and pulse rates measurements.

The potential of IPMCs as sensors has been theoretically expounded particularly in the area of large deformation and bending transduction. Models detailing the swelling dynamics of ionic polymeric gels, the diffusive mechanisms and ion transport involved in polymer bending and actuation, along with expressions for the deflection of IPMCs in an applied electric field have been derived [60]. The interested reader is referred to a review by Shahinpoor and several other studies in this area for details [60,64,65]. The practical performance of IPMC materials in this area, however, has yet to be fully and systematically characterized [57]. The selection of IPMCs for further investigation was motivated by their relatively novel and exploratory status together with the potential that they have exhibited as sensors thus far. Exploration of IPMC sensors for hand prostheses where large deformation bending is expected is therefore a logical step and will be detailed in the following sections where we will demonstrate the high potential of IPMC sensors and ease of implementation.

4. Empirical analysis of IPMC sensors

A gold-coated IPMC film, 0.3 mm in thickness, 3.4 cm in length, and 0.7 cm in width (Environmental Robots Inc., New Mexico, USA) was selected as a suitable geometric fit for the metacarpophalangeal joint of a typical hand prosthesis. The specific objective of this study was to characterize the material response to quasi-static and dynamic bending in the context of a prosthetic hand. To this purpose, the performance of a calibrated IPMC sensor was evaluated in terms of prediction errors for a range of bending rates and angles typical of a prosthetic hand.

4.1. Methodology and apparatus

The IPMC sample was loaded as a cantilever beam and bent from 0 to 90° at varying rates of bend in air without hydration. The apparatus employed for this analysis was designed to emulate, in a controlled environment, the IPMC's potential function as a bending sensor in a prosthetic hand in terms of expected bending rates, angles, and overall sensor dimensions and constraints. In this way, specificity of results to a particular prosthesis was avoided as ideally, sensors of this type could be fitted aftermarket and would not necessarily have to be intrinsic to the initial design of the prosthesis. Specifically, one end of the IPMC sample was fixed in a stationary clamp fitted with isolated electrodes in order to measure the voltage potential across the polymer. The opposite end was fixed to a rotating platform operated by a stepper motor with a step size of 0.9°, and controlled by the computer via the parallel port in order to vary bending angles and rates reliably. The input as measured by the sensor is therefore the angle of rotation of the stepper motor shaft, henceforth referred to as the bending angle, whilst the shaft angular velocity is subsequently denoted as the bending rate. The output voltage was amplified 100 times (Grass 15A54 Amplifier System), converted to a digital signal via a data acquisition board (NI 6014, National Instruments) at

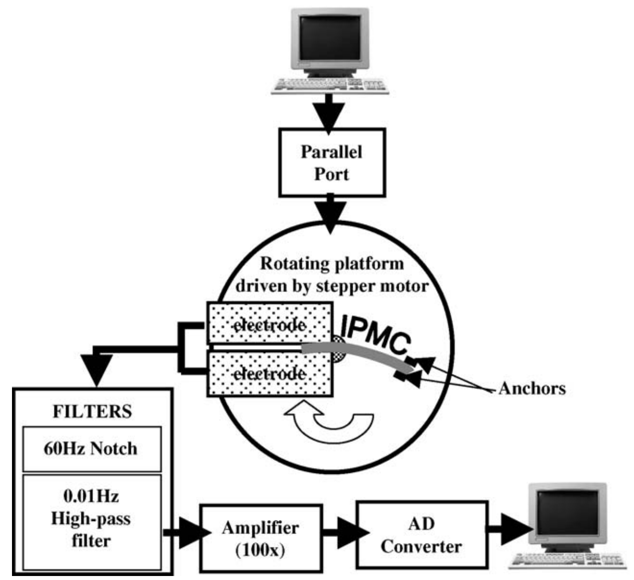

Fig. 1. Schematic of experimental set-up. The polymer is mounted rigidly on one end and anchored to a rotating platform driven by a stepper motor on the other. The polymer response is filtered, amplified, and digitized for signal processing.

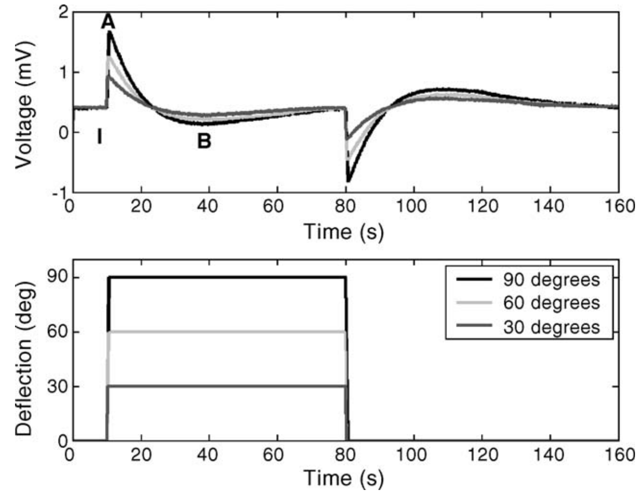

Fig. 2. Gold-coated IPMC voltage response to deflections of varying bending angles. The initial resting voltage (I) is non-zero followed by a voltage spike (A) upon deflection of the IPMC sensor. This voltage spike is followed by a period of adaptation and recovery featuring a voltage trough (B). Similarly, a negative deflection incurs an identical response of opposite sign.

a rate of 1000 samples per second and processed in MatLab 7.0. To reduce signal noise, low (below 0.01 Hz) and 60 Hz frequencies were filtered. The recorded signal was filtered further using a Daubechies 4 wavelet with soft thresholding [66]. Fig. 1 provides a schematic of the experimental set-up.

Using the above-described apparatus, training data were accumulated for sensor calibration. For each training case, the sensor signal was monitored for a duration of 90 s. During the first 10 s, the resting voltage of the IPMC sensor was established after which the polymer was bent at a constant rate (between 10 and 110° s^{-1}) to a specified angle (between 10 and 90°). The polymer was maintained at this angle until the measured potential returned to that of the initial resting voltage. At that point, the IPMC sensor was bent at the same rate to 0°. In this way, motion typical of both flexion and extension was captured. Data collection was manually triggered. The training set consisted of 160 training cases at varying rates and angles, selected to provide equal representation throughout the ranges.

From each of the signals in the training set, the necessary information could be extracted to build calibration curves for both bending rate and angle of bend. Specifically, the peak voltage of the signal was measured (point A in Fig. 2), along with the time required to reach this voltage peak. This time physically corresponded to the duration over which the polymer was in motion. Using these two values, calibration curves, as presented in the subsequent results section, were generated to predict bending rates and angles.

In order to test the calibration and the performance of the IPMC sensor, a test set numbering 100 samples with rates varying from 25 to 120° s^{-1} and bending angles ranging from 10 to 90° was collected. The degree of bend was predicted directly from the developed calibration curve relating the magnitude of the voltage peak to the angle of bend. Two methods were employed to calculate the rate of bend. In the first, herein referred to as the direct method, the rate was predicted simply by dividing the predicted bend angle as determined by the voltage peak by the measured time over which the bend occurred. The second method, dubbed the calibration method, involved the use of a second set of calibration curves which relate the predicted bending angle and the signal slope in the region of the voltage peak (I–A, Fig. 2) to the bending rate. This relationship will be clarified in the ensuing section.

A final experiment was conducted to gauge the response of the IPMC material to higher frequency stimuli. Such stimuli might be realistically experienced by sensors deployed in the typical "cookie crusher" control scheme in which the hand quickly closes in the absence of a control signal from the user. In this experiment, the polymer was subjected to 90° bends at 1 and 0.5 Hz. Furthermore, in order to evaluate the IPMC's ability to distinguish between varying bend angles at higher rates of stimuli, the response of the IPMC material was also observed for bend angles rapidly alternating between 90 and 30°.

4.2. Results

The IPMC is characterized by a non-zero resting potential and exhibits a highly repeatable voltage response to mechanical deflection. A dynamic bending stimulus triggers a voltage response followed by a gradual return to the initial resting voltage in the absence of dynamic deflection. This response, as depicted in Fig. 2, is reminiscent of that of many neurons

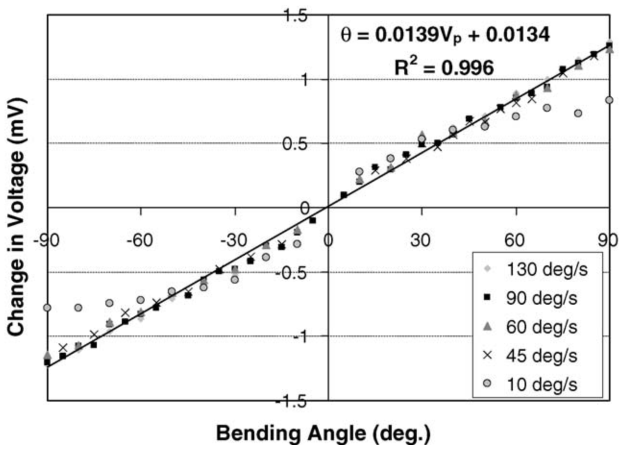

Fig. 3. Peak voltage vs. bending angle at varying bending rates. For bending rates greater than $45^\circ\ s^{-1}$, the relationship is linear with a correlation coefficient of 0.996. At lower bending rates, i.e., $10^\circ\ s^{-1}$, this relationship becomes increasingly non-linear.

when subject to a step input (i.e., a non-zero resting potential followed by an initial response succeeded by adaptation and recovery). Evidently, there are several distinct features of the IPMC response that could potentially be used to distinguish bending rates and degrees of bend.

4.2.1. Bending angle—defining equation

An initial observation indicates that the magnitude and timing of the voltage trough (B) in the adaptation and recovery periods of the polymer response was not correlated to bending rate or bending angle. However, the magnitude of the initial voltage peak (A) was strongly related to the angle of bend applied and its sign was representative of the direction of bend. As depicted in Fig. 3 and defined in (1), a clear linear relationship existed between the angle of bend (θ) and the resulting voltage peak (V_p) at bending rates approximately greater than $45^\circ\ s^{-1}$.

$$\theta = 0.0139\ V_p + 0.0134 \tag{1}$$

At increasingly lower bending rates (i.e., $10^\circ\ s^{-1}$ as depicted by the shaded circular markers in Fig. 3), the relationship between the angle of bend and the resulting voltage peak became increasingly non-linear.

The magnitude of the initial voltage peak was not strongly influenced by the rate of bend. As illustrated in Fig. 4, for small degrees of bend (i.e., <45°), the voltage peak was statistically independent of the rate of bend. This was also true for larger degrees of bend at increasingly high bending rates. This characteristic is useful as it enables the initial voltage peak to be largely decoupled from the rate of bend at rates typical of prosthetic hands, i.e., 45–90° s^{-1} [67]. Thus, within this operating range, demarcated by the vertical dashed lines in Fig. 4, the initial voltage peak is a relatively independent indicator of the bending angle and (1) applies.

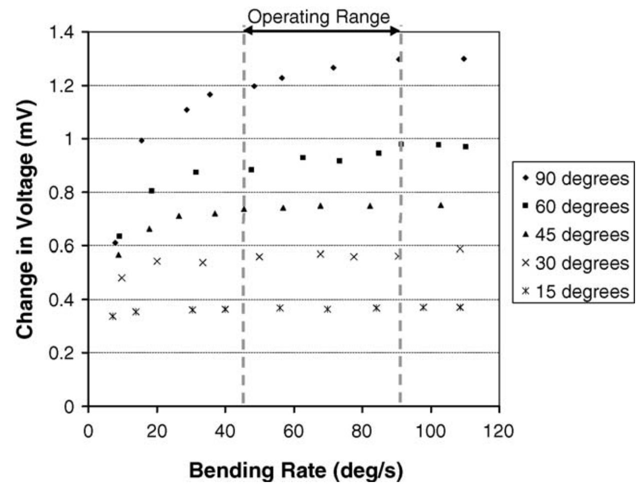

Fig. 4. Peak voltage vs. bending rate for varying degrees of bend.

4.2.2. Bending rate—defining equation

Although the rate of bend was not reflected in the initial voltage peak, it was effectively captured by the slope of the voltage signal in the region of the voltage peak (i.e., from points I to A in Fig. 2). The apparent slope (i.e., the ratio of the peak voltage to the time of bend in the I–A region) exhibited a linear relationship with the rate of bend for all bending angle rates as depicted in the calibration curve, Fig. 5. The slopes of the calibration lines of Fig. 5, herein referred to as the calibration factors, decreased with increasing degree of bend. Fig. 6 depicts the logarithmic relationship between the calibration factor (F_c) and the degree of bend (θ) as defined in (2).

$$F_c = -0.0042\ \ln(\theta) + 0.0325 \tag{2}$$

The calibration factor has units of mV deg^{-1}. From simple dimensional analysis, the bending rate, ω_c, estimated by the

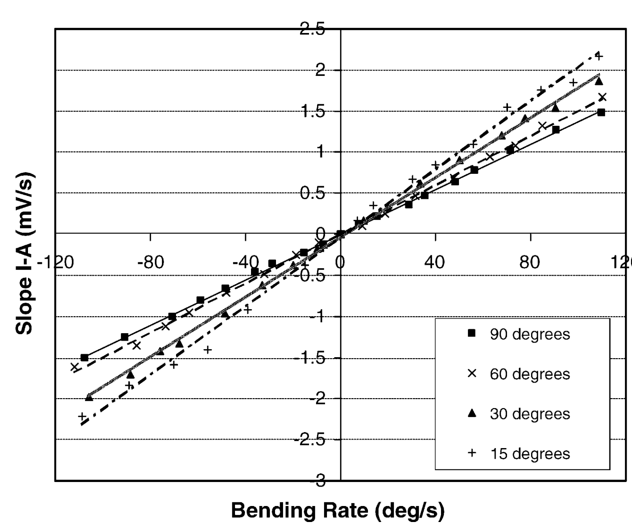

Fig. 5. Linear relationship between the slope (peak voltage divided by time of bend) in region I–A and the bending rate for varying degrees of bend.

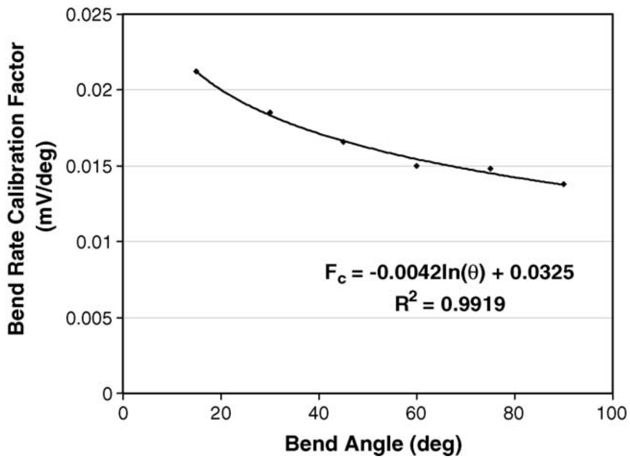

Fig. 6. Logarithmic relationship between the calibration factor and the degree of bend.

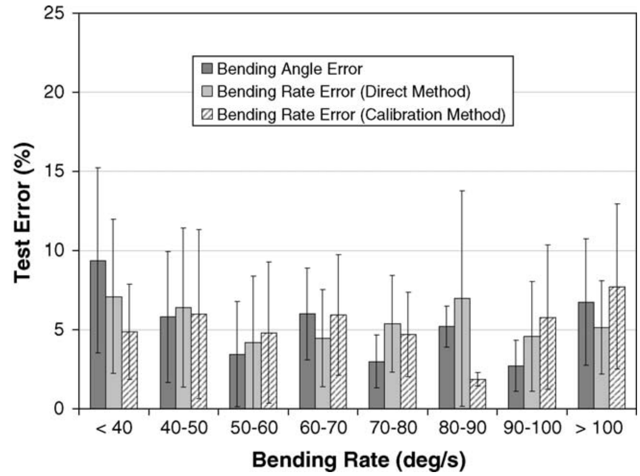

Fig. 8. Test errors for predicted bending angles and rates for various bending rates.

calibration method, was given by:

$$\omega_c = \frac{V_p}{(F_c t)} \tag{3}$$

where V_p is the voltage peak, F_c the calibration factor defined by (2) and t is the total time of bend.

Conversely, the simpler direct method, the equation for which is presented in (4), did not require additional calibration and involved only the bend angle as predicted in (1) and the time of bend in order to predict the bending rate (ω_d).

$$\omega_d = \frac{\theta}{t} \tag{4}$$

4.2.3. Predicted angles and rates

Figs. 7 and 8 depict the calculated percentage error between the predicted and actual values for varying bending angles and bending rates for a test set of 100 samples. The overall percentage errors were calculated for bending rates and bending angles within the typical operating ranges of powered prosthetic hands, namely 45–90° s^{-1} and 10–90°, respectively. An average error of 4.4 ± 2.5% was calculated for bending angle predictions. The bending rate predictions as determined via the direct method exhibited an average error of 5.3 ± 4.3%. Alternatively, the calibration method predicted the rate of bend with an average error of 4.8 ± 3.5%. No statistical difference in average error was observed between the direct and calibration methods ($p = 0.4$), however, the standard deviation was statistically smaller using the calibration method ($p = 0.03$). The direct method for calculating bending rates is extremely vulnerable to errors in the measured time, whereas this susceptibility is limited to a certain degree by the calibration process used in the calibration method. This suggests certain advantages in terms of robustness to the credit of the calibration method for implementation in hand prostheses.

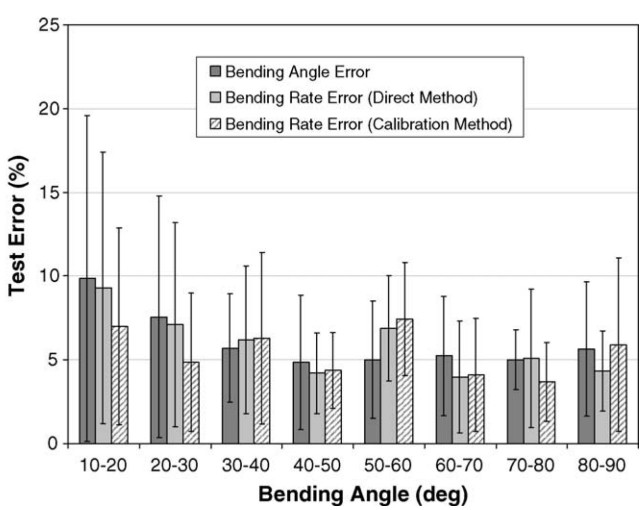

Fig. 7. Test errors for predicted bending angles and rates for various degrees of bend.

As observed in Fig. 8, the magnitudes and deviations of bending angle errors increased significantly for bending rates lower than 40° s^{-1} ($p = 0.002$). Furthermore, as evident in Fig. 7, bending angle errors and standard deviations also tended to increase for decreasingly small deflections with statistically greater average errors for bending angles below 20° ($p < 0.09$) and larger standard deviations for bending angles below 30° ($p < 0.01$). This error may be attributed to the reduced signal to noise ratios observed at smaller angles of bend which produce smaller changes in voltage potentials relative to the resting potentials. The standard deviations of bending rate errors as predicted by the direct method increased significantly for angles of bend less than 30° ($p < 0.03$), while those of the calibration method exhibited a marked increase for angles below 20° ($p < 0.04$). In both cases, there were no statistically significant trends of note with regards to the average bending rate errors.

4.2.4. High frequency stimulus

Fig. 9 presents a final example of the response of the IPMC material to higher frequency stimuli as might be experienced

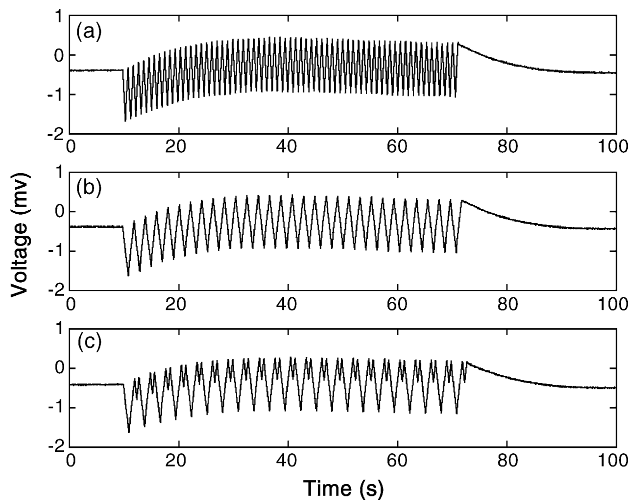

Fig. 9. IPMC response to (a) 1 Hz, 90° bending stimuli; (b) 0.5 Hz, 90° bending stimuli and (c) an alternating 90 and 30° stimuli.

in prosthetic hands. Fig. 9a presents the IPMC's response to a 90° bend cycle at a frequency of 1 Hz, while Fig. 9b depicts its response to a 0.5 Hz stimulus. Fig. 9c exhibits the polymer response to alternating bends of 90 and 30°. In all cases, the degree of bend is accurately captured by the relative change in voltage. Interestingly, a gradual increase in signal magnitudes is observed at the onset of the stimulus in all three cases. The polymer response to higher frequency stimuli is therefore characterized by both a transient and a steady state region. The mechanisms responsible for the electromechanical response of IPMC materials are not yet fully understood. Explanations regarding such transients remain speculative. As evident from a comparison of Fig. 9a and b, the relative change in voltage for a given degree of bend was not dependent on the stimulus frequency within the range of 0.5–1 Hz as typical of prosthetic hands.

5. Discussion

By studying the voltage response of the IPMC polymer, critical information regarding the bend rates can be simply and reliably extracted. IPMC polymers are dynamic sensors and do not maintain a constant voltage response for static deflections. Angles and rates of bend therefore cannot be determined by absolute voltage readings, but must be obtained from changes in voltage potentials. In parallel, the sensor does not measure the absolute angle of bend, but changes in the angle of bend. The controller must therefore incorporate the requisite "memory", determining the present bend angle based on the change in voltage potential produced by the sensor, in addition to the previous angle of bend. The information necessary to accomplish this is collected entirely during the dynamic phase (i.e., I–A in Fig. 2), therefore transient outputs typical of the adaptive phase (i.e., A–B in Fig. 2) are immaterial.

5.1. Practical challenges

In contrast to the advantages of form and sensitivity offered by IPMC polymers, there are also several additional challenges, which are not encountered in the use of more traditional sensors. Firstly, IPMC polymers are responsive not only to bend, but also to pressure and stretch. Therefore, the implementation of such materials in the mechanical design of a prosthetic hand must ensure that the sensor is isolated from these forms of stress. Secondly, IPMC polymers may respond to varying temperatures. When activated in air, a temperature dependent response was not observed between temperatures ranging from 5 to 50 °C, however, further characterization of the temperature response of IPMC materials should be conducted to ensure reliable operation within the range of operating temperatures typical of a hand prostheses. The response of these materials is also highly dependent on their moisture content, which may complicate calibration. For sensing in air, as presented in the above results, signals on the order of millivolts are produced. No discernable change in the polymer response was observed over a 3 month period during which the polymer was exposed to air. In wet environments, however, the response of the IPMC sample jumped an order of magnitude to tens of millivolts. The thresholds and resolution of these sensors therefore depend greatly on the moisture content. In addition, thresholds of detection also depend on the rate of bend. For example, when activated in air, the IPMC polymer remained insensitive to large angle deformations at bending rates of less than about 0.75° $\min^{-1}$ (Note: this is comparable to the threshold sensitivity of the natural proprioceptive system, i.e., 1–4° $\min^{-1}$ [7]). Conversely, very small (i.e., 1°), but rapid deflections, were reliably detected. The IPMC polymers therefore respond much in the manner of the RA afferents innervating the hand, both in terms of sensitivity to dynamic stimuli and the phasic nature of the response. Lastly, the general characteristic response of IPMC materials has been well-established (i.e., a linear voltage response to deflection) both in this paper and in previous study [68]. The quantitative measures (i.e., the slope associated with the linear relationship), however, depend largely on the manufacturing techniques and standards employed [59,62]. Calibration is therefore a necessary and important practice in the use of IPMC sensors.

5.2. Filling the void

A distinct void in the availability of light and reliable bending sensors for proprioceptive applications currently exists. Reliable technologies such as strain gauges are typically not very durable, while fibre optics require additional light sources and added complexity, and magnetic-based sensors such as those utilizing the Hall effect can become bulky when used for bend detection [69]. Typically, selection is therefore limited to resistance-based bend sensors, many of which are not very reliable as determined by a study by Simone and Kamper [69] in which the most promising bend sensor exhibited a highly non-linear response with repeatability errors ranging from 1.9 to 5.4% for a single grip configuration. Based on these findings, the linear response of the IPMC sensor, its ability to detect bend rate as well as bend angle, and its comparable error rates, suggest its promise as a proprioceptive sensor for prosthetic applications.

5.3. Biological analogues: EAP potential in hand prosthetics

A deeper appreciation of the intricacies of the biological sensory system of the hand yields valuable insight as to the design of its engineering counterpart. Although the day when engineered devices will paint a sensory picture on par with that of the biological hand remains for the moment a topic of science fiction, several permeating features of the natural sensory system could and should be incorporated into engineering designs. Firstly, the importance of form to function should be underlined. The location and structure of the sensory receptors in the natural hand are highly variable and key to their underlying function [70,71]. In parallel, engineered sensors ranging in form and structure are also necessary to capture the varying functional tasks required by the prosthetic hand, from proprioception to slip control. Electroactive polymers offer this potential and are easily manufactured in a range of forms from fibres, to films, to fabrics or strips. Furthermore, the proven potential of EAPs in MEMs technology [56] promises an age of highly miniaturized sensors with improved resolution and ease of implementation. This may further encourage the use of sensor arrays. In the natural hand, sensory information is often processed in terms of a population of receptors in order to extract additional information beyond that offered by individual receptors [1,70]. This principle has been adopted in endeavours toward the creation of "sensitive" skin whereby higher level processing of arrays of force sensors lend information regarding the motion, direction, and orientation of stimuli as well [33].

A second feature of the natural hand is the concept of specificity, exemplified by the varying filter properties exhibited by the different sensory receptors [4,70]. Each of the four groups of mechanoreceptors for instance, is largely responsible for a dominant sensory function and responds within a given range of stimuli [70,72]. Additionally, the manner of response (i.e., tonic or phasic) is well-suited to the stimulus. Likewise, engineered sensors must also maintain this specificity and respond in a predictable manner within a practical range of operating conditions. This may present a degree of challenge to the use of some varieties of ionic EAPs which are highly sensitive to moisture content for example. Efforts to overcome this limitation are ongoing including the development of protective coatings that limit dehydration. On the other hand, the diversity of EAPs offers a unique range of responses paralleling that of the natural hand. As presented, EAP candidates for detection of bending, strain, vibration, and pressure are available with varying responses mirroring both the slowly and rapidly adapting afferents innervating the natural hand. This diversity, in conjunction with the ability to tailor the chemical composition of these materials in order to tune their sensory specificity and response, presents them as viable candidates for engineered sensors in hand prostheses.

A third general feature of the natural hand is its capacity for information integration and processing [1,70]. With tens of thousands of receptors, the natural hand produces vast amounts of sensory information, which is efficiently transformed to yield an accurate sensory picture of our external environment. Electroactive polymers hold much promise in satisfying the need for suitable sensors, but they have also been proposed as a mode of communicating sensory information to the user, an important challenge in the design of sensory systems. Electroactive polymers enable the conversion of ionic signals, typical of the membrane potentials developed at the neural level, to electronic signals and vice versa. Development of electroactive polymers suitable for neural interfaces is currently underway in the hopes of providing a connection to the peripheral nervous system through which sensory information and user response could be conveyed [73].

6. Conclusions

In conclusion, electroactive polymers are a novel class of material whose full potential has yet to be realized. The primary advantages of EAPs over more traditional sensing techniques for implementation in hand prostheses lie in their flexibility, manufacturability, diversity, and dual role as actuator and sensor. The state of technology is at varying stages of development ranging from well-established conductive rubber pressure sensors, to the more revolutionary ionic metal polymer composites. In this study, we have analyzed the functional potential of IPMC sensors for measuring bending angles and rates in the context of hand prosthetics. Average IPMC sensor errors were on par with those of the natural proprioceptive system, for which mean amplitude errors of 3–5° are typical in the metacarpophalangeal joints [7]. These errors were also comparable to conventional resistance-based bend sensors with the added benefit of a highly linear response. Conventional sensors, such as fibre optics, strain gauges, and magnetic-based sensors also offer high resolution and accuracy, however, without the required durability, flexibility, and compact size, it is improbable that they will be incorporated into clinically acceptable prosthetic devices. Contrary to many traditional approaches, IPMCs also do not require external power supplies or auxiliary mechanisms, such as light or magnetic sources which add bulk, energy requirements, and complexity to the system, making the incorporation of such sensors aftermarket impractical, if not impossible. In a practical prosthetic controller, the recommended signal filtering employed in this study for an IPMC sensor, could be achieved with a combination of digital processing on a microcontroller platform and analogue manipulation using small, low power integrated circuits. Their unique attributes and their favourable response as demonstrated in this study endorse IPMC sensors as a viable and much needed alternative if proprioceptive sensory feedback is to be realized in prosthetic applications. Electroactive polymers may indeed provide the jump-start needed to evolve the next generation of prosthetics with promises of increased functional and sensory capacity.

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