Energetic Studies of Emaciated Film based Sensor for Biomedical Applications

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Abstract

The state-of-the-art in CP-based sensing devices and CP-based sensor elements that may be used as instruments for surgical procedures and clinical diagnosis are the main topics of this article. A brief discussion is held on three examples of CP-based sensor applications: temperature sensors, tactile sensing "skins," and electrochemical biosensors. Lastly, a few of the most important problems with CP-based sensors are outlined.

The fundamental building blocks of life, proteins, nucleic acids, and polysaccharides, as well as commercial goods derived from the automotive, construction, and transportation industries, plastic toys and tools, reading glasses, and other items, are all composed of polymers (Carraher, 2010). To create polymer composites, the majority of these compounds combine one or more different components.

Keywords: Energetic, Emaciated, Film, Sensor, Biomedical, Application. Popular, Properties.

• Introduction

A class of organic polymers, known as conducting polymers (CPs), has become increasingly popular due to its unique electrical and optical properties. Material characteristics of CPs are similar to those of some metals and inorganic semiconductors, while retaining polymer properties such as flexibility, and ease of processing and synthesis, generally associated with conventional polymers. Owing to these characteristics, research efforts in CPs have gained significant traction to produce several types of CPs since its discovery four decades ago. CPs are often categorised into different types based on the type of electric charges (e.g., delocalized pi electrons, ions, or conductive nanomaterials) responsible for conduction. Several CPs are known to interact with biological samples while maintaining good biocompatibility and hence, they qualify as interesting candidates for use in a numerous biological and medical applications. Naturally occurring polymer composites such as bone (combina- tion of ceramic calcium phosphate crystallites and collagen fibres

1. Representative applications

1.1. *Electrochemical sensors*

Clark and Lyons (1962) developed the first biosensing device almost 50 years ago by integrating enzyme and glucose oxidase to an electrode. Since then, much progress has been made in the development of biosensors for use in diagnostic detection and monitoring of biological metabolites. Moreover, recent advances in lab-on-chip devices have stimulated demand for portable, highly sensitive and precise analytical tools for easy and real-time estimation of desired analytes such as glucose, cholesterol, anti- bodies, nucleic acids, hormones, drugs, viruses, neurotransmitters, pathogens and toxins. An electrochemical biosensor typically con- sists of a sensing element and a transducer (Gerard et al., 2002). The sensing element is a biorecognition layer made up of biomolecules (e.g., enzymes act as biorecognition entities in an enzymatic biosen- sor) that interacts with the analyte of interest producing a chemical signal detectable by the transducer, which ultimately transforms the input to give an electrical readout.

Since the discovery of metallic polymer (Shirakawa et al., 1977), conducting polymers (CPs) have been extensively used as trans- ducers in electrochemical biosensors to measure and amplify signals (Cosnier, 2005). Both intrinsically conducting polymers and conducting polymer-nanocomposite materials have been used as bio-transducers. Carbon nanotubes (Liu et al., 2006; Perˇıez et al., 2005; Pumera et al., 2006; Wang and Musameh, 2003) and metal nanoparticles (Park et al., 2004; Yu et al., 2003; Zeng et al., 2009; Zou et al., 2010) are some of the most commonly used filler mate- rials for polymer composite-based functionalization of electrodes in electrochemical sensors.

Some of the most commonly used ICPs for development of different types of electrochemical biosensors are: PANI, PPy and PT. Low cost, scalability, easy processing capa- bility and material properties such as large surface area, adjustable transport properties, and chemical specificities makes conducting polymers attractive candidates for applications in electrochemical sensing (Sree et al., 2002). Conjugated CPs contain alternating single and double bonds in their polymer chain resulting in the formation of de-localised electrons which act as charge carriers. In order to improve sensitivity and selectivity of the biosensors, redox media- tors are dispersed, added as dopants or chemically conjugated into the polymer matrix (Chen et al., 2004; Cosnier et al., 2003; Fiorito and Brett, 2006). Conjugated CPs, thus, mediates electron transfer between the biorecognition layer and the final electrode (Gerard et al., 2002). Moreover, the conjugated backbone of CPs allows modulation of its properties by enabling attachments (or immo- bilization) to a variety of chemical moieties. For example, Fig. 1 shows an illustration of avidin-functionalized PPy nanowires used for sensing biotin-conjugated DNA molecules (Ramanathan et al., 2005). Electrodeposition procedure based on polymerized films is a common technique used to immobilize macromolecules. Highly reproducible, ultrathin layers of CP coatings can be achieved by this technique. Furthermore, depending on the type of polymer,

emerged opening up exciting new applications in several fields including bioelectronics. The CPCs typically consist of a combina- tion of one or more non-conducting polymers and conductive-filler materials distributed throughout the polymer matrix. The conduc- tivity of CPCs is governed by percolation theory, which describes the conductive phase of CPC formed by a network of the filler materials at a given weight percentage. For filler loadings below a certain con-centration, the filler particles no longer maintain physical contact with each other to provide continuous path for electron trans- port necessary for conduction and subsequently, the CPC exhibits a sudden drop in conductivity (percolation threshold). Some of the drawbacks of CPCs include high dependence on processing condi- tions, mechanical instability and an insulated surface layer over the conducting material (Freund and Deore, 2007).

Metallic conductivity in organic conducting polymers such as crystalline polyacetylene films combined with *p*-type dopants (oxi- dants) was first discovered by Shirakawa, MacDiarmid and Heeger in 1977 (Chiang et al., 1977; Shirakawa et al., 1977). Soon, *n*-type dopants (reducing agents) were found to depict similar effects (Chiang et al., 1978). Following these discoveries, a new class of organic conducting polymers, also known as intrinsically con- ducting polymers (ICPs), was established. ICPs contain monomers capable of acquiring positive or a negative charge through oxidation or reduction which in turn contributes to the electrical conductivity in ICPs. Some examples of ICPs are polyacetylene (PA), polypyrrole (PPy), polythiophene (PT) and polyaniline (PANI). Two other classes of conducting polymers that emerged around the same time are: redox polymers and ionically conducting polymer (polymer/salt electrolytes). Redox polymers are less conductive compared to ICPs. They have localized electron redox sites that contribute to the elec- trical conductivity. In ionically conducting polymer, as the name suggests, conduction is achieved through flow of ions. Their use in electrochemical sensing is largely limited by the low ionic con- ductivity at room temperature and time-dependent increase in resistance of the polymer electrolyte (Freund and Deore, 2007). ICPs have highly flexible chemical structure that can be modified to acquire desired electronic and mechanical properties. Since the ICPs have the ability to efficiently transfer electrons produced by biochemical reactions, they have been used extensively in biosen- sors in the form of transducer that form an intermediate layer between biological samples and the electronics used for signal readouts. They are also known to be compatible with biological molecules in neutral aqueous solution. For the same reason, ICPs have attracted much attention as a suitable matrix for entrapment of biomolecules. Several studies have explored these unique mate- rial properties of ICPs to produce a wide range of biosensors for measurement of vital analytes relevant to clinical diagnosis (Bidan et al., 1988; Borole et al., 2006; Boyle et al., 1989; Cosnier, 1999; Gerard et al., 2002; Janata and Josowicz, 2003; Kranz et al., 1998; Lewis et al., 1999; Schuhmann, 1995; Trojanowicz et al., 1997). Blends of ICPs and CPCs have also been investigated in order to improve mechanical stability and processability of CPCs (Freund and Deore, 2007).

In this paper, we review the state-of-art of conducting polymer-based sensors developed for biomedical applications. The sensor type depends on the parameters-of-interest such as skin/tissue temperature, force exerted by tissues/blood vessels during surg- eries, and the presence of biochemical components like glucose and cholesterol. CPs possess excellent electrical, chemical and mechan- ical properties useful for designing efficient, real-time and versatile biosensors. Several reviews and studies based on specific type of sensing mechanism employed by one or more classes of CP, or vice-versa can be found in the literature (Bidan et al., 1988; Borole et al., 2006; Boyle et al., 1989; Cosnier, 1999, 2007; De Rossi et al., 2005; Gerard et al., 2002; Janata and Josowicz, 2003; Kaushik et al., 2008; Kranz et al., 1998; Lewis et al., 1999; Mueller,

1.2. *Tactile sensor (artificial skin)*

MEMS-based tactile sensors have tremendous potential for applications in medical robotics, interactive electronics, and telemedicine. Robust, reliable and real-time haptic feedback pro- vides a range of tactile information such as tissue compliance, texture, contact forces and torques, dynamic slip sensing, and pressure-distribution useful for identification, localization and monitoring of critical anatomical structures. The advent of min- imally invasive surgery (MIS), which is the use of specifically designed surgical instruments and visual devices that allow surg- eries to be performed through small incisions, offers distinct advantages to the patients in terms of reduction in intra-operative blood loss, risk of post-operative infection, less traumatic surgery and accelerated recovery. Since the MIS operating field cannot be directly accessed by surgeons, there is an increasing need for inte- gration of tactile sensory devices onto the surgical tools used in MIS. This has allowed researchers to adopt an interdisciplinary approach towards developing new techniques and technologies to overcome the inherent drawbacks involved in the MIS procedures (Schostek et al., 2009). The key idea behind integrating tactile feed- back (or haptic interface) onto the surgical probes is to increase the effectiveness of the surgery by allowing the surgeons to measure variations in the superficial tissue properties such as temperature, texture and contact force, to feel the hardness or tension of tissues, and to evaluate anatomical structures such as nerves, vessels and ducts (Puangmali et al., 2008).

Tactile sensing provides improved dexterity, dynamic gripping and manipulation by robots and humans (Lee and Nicholls, 1999). In the past two decades, tremendous efforts have been towards developing a human-skin-like sensor that can potentially provide a broad spectrum of tactile information particularly useful in the field of medical robotics and certain surgical procedures (Beebe et al., 1995; Engel et al., 2003a,b; Hu et al., 2007; Jiang et al., 1997; Kane et al., 2000; Kolesar and Dyson, 1995; Reston and Kolesar, 1990; Sekitani et al., 2008; Shimizu et al., 2002; Someya et al., 2005; Yang et al., 2008). Different types of materials used for tactile sensing includes silicon-based piezoresistive (Beebe et al., 1995; Kane et al., 2000) or capacitive sensors (Gray and Fearing, 1996; Leineweber et al., 2000), and polymer-based piezoelectric, capaci- tive or piezoresistive sensors (Kolesar and Dyson, 1995; Reston and Kolesar, 1990). Recently researchers have explored the possibility of using composite-material sensors by combining both silicon and polymers, examples of which includes embedding of silicon sens- ing elements in polymer skins (Beebe and Denton, 1994; Jiang et al., 1997; Wen and Fang, 2008), packaging of silicon-based sens- ing devices in protective casing of polymer layer (Gray and Fearing, 1996; Leineweber et al., 2000; Kane et al., 2000), etc. Silicon-based tactile sensors have proven to provide high sensitivity, high spatial resolution and ease of integration into electronic devices. However, the brittleness of the silicon materials limit its use as a flexible or stretchable tactile sensor particularly when the sensors need to be packaged onto curved surfaces of surgical probes or robotic manipulators. Furthermore, the finite size of silicon wafers imposes size-related design constraints on the dimensions of the tactile sen- sors. Polymer-based tactile sensors, on the other hand, are more flexible, not limited by dimensions, and chemically resistant (Liu, 2007).

The use of a piezoelectric polymer, polyvinylidene fluoride (PVDF), for tactile sensing was first reported by Dario and de Rossi (1985) which was soon followed by several other studies using PVDF or their copolymers for tactile sensing (Choi et al., 2005; Dargahi et al., 2000; Kolesar et al., 1992, 1996; Yamada et al., 2002;

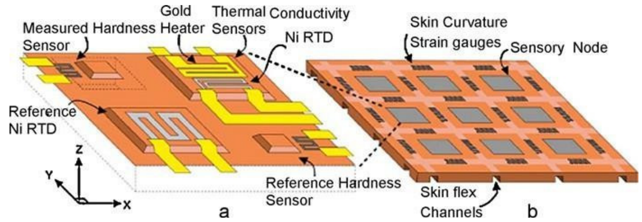

Fig. 3. A schematic of the multimodal tactile sensor. (a) A sensory node incorporates 4 distinct sensors: a reference temperature sensor, a thermal conductivity sensor, and contact force and hardness sensors. (b) Sensor nodes are arranged in an array to form skin. (*Source*: Engel et al., 2003a,b). Yuji and Sonoda, 2006). De Rossi et al. (1993) reported a piezoelec- tric polymer-based, skin-like tactile sensor which was selectively sensitive to stress and shear forces. Thin polyimide film (Kapton) was used as a supporting structure, and PVDF and polyhydroxybu- tirate (PHB) were used as sensor elements. Dargahi et al. (2007) reported an experimental design and a theoretical model of an endoscopic tooth-like tactile sensor capable of measuring compli- ance of a contact tissue/object. The sensor set-up consisted of a pair of rigid and compliant cylinders, and two PVDF films. The compli- ance of the object in contact with the sensor was detected by the relative deformation of the rigid and compliant sensor elements. Both the force applied as well as the compliance of the tissue/object sensed can be measured using their sensor. The proposed applica- tion of these sensors is in the field of robotic surgery wherein the sensors can be integrated onto endoscopic graspers (Dargahi et al., 2007).

Two popular mechanisms employed in piezoresistive-based tactile sensing can be broadly classified into: resistive-metal based sensing, and conductive-polymer based sensing. Typical approaches for resistive-metal based sensors involve a flexible array of tactile sensors with each sensory element consisting of piezoresistive metal capable of sensing stress/shear forces (Engel et al., 2003b; Hwang et al., 2007). Inspired from the functionalities of biological skin, Engel et al. (2003b) reported the development of the first "smart skin" based on thin-film piezoresistive-metal-based sensors on a flexible polymer substrate. They developed an array of multimodal tactile sensors capable of sensing contact forces, rela- tive hardness, thermal conductivity and temperature (Fig. 3). The sensing skin consists of an array of sensor nodes arranged on a flexi- ble polyimide (Kapton) substrate. Each node comprised of 4 sensing elements: a thermal conductivity measurement unit, a tempera- ture measurement unit and 2 membranes with metal strain gauges capable of measuring hardness and contact forces. The same group later demonstrated that their multimodal sensor was able to mimic some functionalities of the human skin by identifying objects based on texture classification and other sensory data (Engel et al., 2003b). Although metal-based tactile sensors, in general, give good sensi- tivity and reliable response, relatively complex micromachining is involved in fabricating the sensor elements (Cheng et al., 2010).

A typical conductive polymer-based tactile sensor consists of a flexible, conductive gel or elastomer capable of sensing tactile information, and a set of patterned electrodes for sen- sor readout. Papakostas et al. (2002) proposed a large area force sensor fabricated by sandwiching semi-conductive film in between screen-printed traces of Ag-filled thermoplastic poly- mer. Shimojo et al. (2004) reported fabrication of flexible tactile sensing array using a conductive rubber capable of sensing pressure-related deformations. Thin metal wires that acted as sens- ing electrodes were stitched on the conductive polymer matrix. Someya et al. (2005) developed a flexible, net-shaped pressure and temperature sensor fabricated by processing polyimide or poly(ethylenenaphthalate) and organic transistor-based electronic

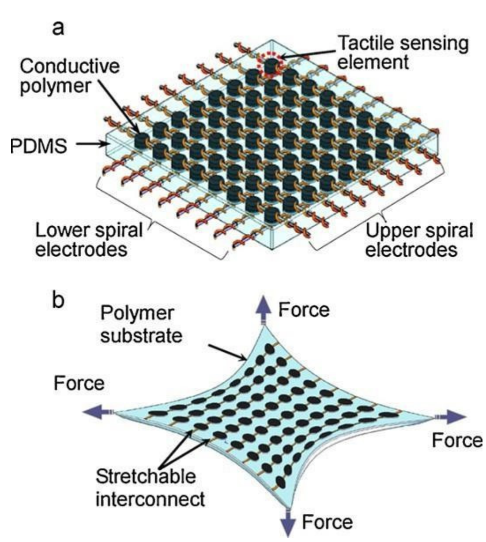

Fig. 4. (a) The schematic of the proposed tactile sensing array. (b) The proposed device under stretching. (*Source*: Cheng et al., 2010).

circuits. Cross-talk between the sensing elements poses a problem for these kinds of sensors. Yang et al. (2008) effectively eliminated cross-talk between adjacent sensing elements (polyimide-copper composite) by dispensing them on a grid of copper electrodes. Tactile devices fabricated using these mechanisms became quite popular because of their durability and ease to manufacture them. Thin metal traces that act as sensing electrodes in polymer-based tactile sensors are mostly vulnerable to large deformations espe- cially caused when the sensor unit are required to be integrated onto complex surfaces. Consequently, research efforts are driven towards increasing the reliability and robustness of these traces. Hu et al. (2007) reported a flexible sensor capable of sensing force and temperature simultaneously. The sensor comprised of a composite material (multi-walled carbon nanotubes dispersed in polydimethylsiloxane (PDMS)) and liquid metal interconnects. Sekitani et al. (2008) proposed a stretchable, elastic conduc- tive material that could be uniaxially and biaxially stretched by 70% without causing mechanical or electrical damage. The sen- sor material was developed by coating PDMS-based rubber on a single-walled carbon nanotubes composite film. Cheng et al. (2010) presented a highly twistable artificial skin by dispensing conductive polymer on a grid of sensing electrodes. The conductive poly- mer was a composite material which is a blend of PDMS and a variety of conductive filler materials (copper, carbon black, cyclo- hexane and silver particles). The sensing electrodes were obtained by winding copper electrodes around nylon fibres. Fig. 4 shows a schematic of artificial tactile skin and the arrangement of the spiral electrodes. The authors demonstrated that the sensor could be twisted up to 70 degrees without any structural or functional damage.

1.3. *Thermal sensors*

Thermal readings have been used in medicine for several cen- turies. In 400 BC, the Greek physician Hippocrates wrote, “In whatever part of the body excess of heat or cold is felt, the disease is there to be discovered”. A common example of this phenomenon is fever, wherein the body temperature is elevated due to circulat- ing pyrogens produced by our immune system. With the advent of new technologies, thermal sensing has become a useful diagnos- tic tool in applications such as thermographic imaging or infrared thermal imaging used for detecting small temperature changes due to vascular disorders, for pre-clinical diagnosis of breast cancer, to identify neurological disorders and monitor muscular perfor- mances (Bagavathiappan et al., 2008). Thermal sensors integrated into tactile sensing probes or catheters has the potential to be used as surgical tools that help clinicians to quantify sensitive changes in tissue temperature during surgical interventions typically in some ablation procedures (removal or abrasion of faulty tissues) that involve heating by laser or RF energy. Moreover, probe-based ther- mal sensors may also provide real-time temperature profiles of tissues which in turn may allow clinicians to precisely control the heat energy, preventing undesired tissue damage.

In order to mimic the thermal sensing and regulation function- alities of biological skin, temperature sensitive transducers used in electronic skins (*E*-skins) generally exploit two types of phys- ical effects: thermoelectricity and pyroelectricity. Thermoelectric temperature sensors, also known as thermocouples, generate elec- tricity from a temperature gradient according to Seebeck effect. A thermocouple refers to the junction between two metals, or semiconductors (*p–n* couples) that is capable of producing volt- age difference relative to a temperature difference. Organic and inorganic materials or combinations of the two have been used as thermocouples (De Rossi et al., 2005). Intrinsically conducting polymers (Feng and Ellis, 2003; Kemp et al., 1999; Morsli et al., 1996; Yan et al., 2002) and carbon/polymer composites (Chung and Guerrero, 2002; Chung and Wang, 1999; Lin et al., 2006) are the most popular organic materials in which thermoelectric property have been studied (De Rossi et al., 2005). Thermoelectric mate- rial parameter, proportional to its efficiency of as a thermoelectric couple, given by the figure of merit, Z, is defined as:

$$Z = \frac{\sigma S^2}{\kappa}$$

σ and $\varkappa$ are the electrical and thermal conductivities respectively, and S is the Seebeck coefficient. The thermoelectric property is more commonly expressed as dimensionless figure of merit: ZT where T is the absolute temperature. Conductive polymer compos- ites have been reported to sense temperature changes based on changes in resistance of the material in a way that the conductivity pathway of the filler elements are affected by the thermal expan- sion and contraction of the polymer matrix. Feng and Ellis (2003) showed that

conjugated nanocomposite polymers can produce a ZT value of ∼1 comparable to commercially available semiconductor- based thermoelectric. For applications in tactile sensing, Someya et al. (2004) fabricated conformable, flexible networks of pressure and thermal sensors using active matrices of organic transistors. In order to develop efficient and stable thermoresistive sensors from conductive polymer composites, the filler material (conductive par- ticles or ICPs) must form a conductive network within the insulating polymer matrix (Feng and Ellis, 2003). To achieve an optimal con- ductive network of the filler material, Nocke et al. (2009) proposed the use of dielectrophoresis for aligning tellurium nanorods in a matrix of poly(vinyl acetate) (PVAc) and a novolak-based posi- tive photoresist. Moreover, they presented photolithographic and stamping techniques for development of resistance-based thermal sensor elements from the nanocomposite material of PVAc and photoresist. Shih et al. (2010) fabricated thermoresistive sensor arrays by dispensing a graphite-PDMS composite on flexible poly- imide films (Fig. 5). The sensor was designed for use as an E-skin to provide haptic interface to robots.

Pyroelectric materials respond to changes in tempera- ture resulting in a spontaneous polarization of the material. Temperature-dependent polarization slightly modifies the posi- tions of the atoms within the crystal and produces a voltage across the material. This change in polarization with respect to tempera.

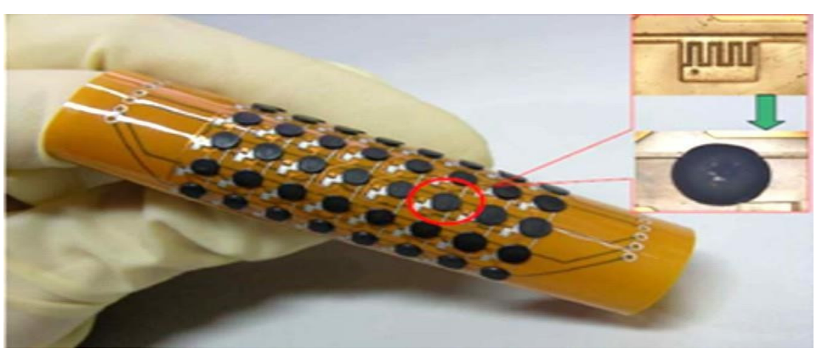

Fig. 5. Flexible temperature sensor array. The insets show the interdigitated elec- trode and composites on the electrode, respectively. (*Source*: Shih et al., 2010).

ture can be described as:

$$p = \frac{\partial P_{S,i}}{}^{\ i} \ \ \partial T$$

where p_i (Cm^{-2} K^{-1}) is defined as the pyroelectric coefficient. Although the pyroelectric effects had been observed and studied for several centuries, investigations of such effects in polymers are relatively new (Bauer and Lang, 1996). Bergman et al. (1971) discovered a strong pyroelectric effect in PVDF, an organic poly- mer, shortly after the discovery of strong piezoelectric property in the same material by Kawai (1969). Based on the pyroelectric properties, Glass et al. (1971) and Yamaka (1972) were among the first to use PVDF for infrared thermal sensing. To date, several studies have investigated the pyroelectric properties of PVDF and P(VDF-TrFE) for applications in thermal radiation sensing (Bauer and Lang, 1996; Hammes and Regtien, 1992; Navid et al., 2010; Setiadi et al., 1999a,b). They typically exhibit pyroelectric coeffi- cients of about 25 μC/m^2 K and 40 μ C/m^2 K respectively which is about 10 lower than the coefficient (380 μC/m^2 K) measured for one of the most commonly used class of ceramics: lead zirconate titanate ($Pb(Zr,Ti)O_3$) (PZT) (Bauer and Lang, 1996). Researchers have, therefore, explored the possibilities of using composite mate- rials to take advantage of the high flexibility of polymers and high performance of ceramics (Malmonge et al., 2003; Sakamoto et al., 2001, 2002).

Polymers have high coefficient of thermal expansion as com- pared to metals and semiconductors. Most polymers absorb infrared radiation because of the vibrational resonance modes present in their organic bonds. Temperature-induced morphologi- cal changes such as conformational changes of polymer molecules and/or rearrangements in their crystal structure may also result in additional energy transduction (Mueller, 2007). Setiadi et al. (1999a,b) developed a polymer-based pyroelectric infrared sensor. The sensor comprised of a conductive polymer (PEDOT:PSS) as an absorber layer and front electrode, a pyroelectric material (PVDF film) and a nickel–aluminium (Ni–Al) metal film as a reflector layer and rear electrode (Fig. 6). The practical use of metal-polymer com- binations of pyroelectric sensors is hindered by the poor adhesion between front electrode (metal film) and sensing layer (polymer- based pyroelectric film). Setiadi et al. (1999a,b) used a conductive polymer (PEDOT:PSS) for effective adhesion of the front electrode to the sensing material (pyroelectric PVDF) film beneath it. The measured IR response was shown to be 10× higher than that of commercially available PVDF films with Ni-Al front and back electrodes.

Several studies have investigated thermomechanical transduc- tion in polymers in view of mimicking thermal IR sensors found in nature. The ability of a jewel beetle (*Melanophila acuminata*) to detect forest fires from a distance of 60–100 miles has been contributed to the alternating hard and compliant nanolayers of microthermal sensors found in bilateral thoracic pit organs of the beetle (Campbell et al., 2002; LeMieux et al., 2006). Based on the principle of thermomechanical transduction, LeMieux et al. (2006) developed a bimaterial polymer-silicon microcantilever with a temperature resolution of 0.2 mK and thermal sensitivity of 2 nm/mK. The high thermal sensitivity was attributed to the strong thermal stress induced by the plasma-enhanced polymeric nanolayers adhered onto the silicon substrate of the microcan- tilever. Inspired from the biological IR sensors, Mueller (2007) investigated the effect of changes in temperature on chitosan, a deactylated form of a biomaterial (chitin) found in the sensory organs of the jewel beetle. As an initial prototype, the author fab- ricated bimorph microcantilever from novolak-resin photoresist and polysilicon for IR thermal sensing. Lin et al. (2006) devel-oped a thermally sensitive microcantilever using a layer-by-layer structure of conductive polymer, metal and ceramic. The trilay- ered microbeam comprised of silicon nitride, a ceramic with low thermal expansion, an ultrathin film of gold followed by a top- most layer of chemically grafted polymer brushes (Fig. 7). The nanocomposite polymer structures of functionalized polyacryloni- trile (PAN) and polystyrene (PS) were reinforced with silver (Ag) nanoparticles and single-walled carbon nanotube bundles in order to enhance the mechanical strength of the composite polymer layer. The microcantilever design uses the phenomenon of temperature- induced bending of layered beams composed of one or more

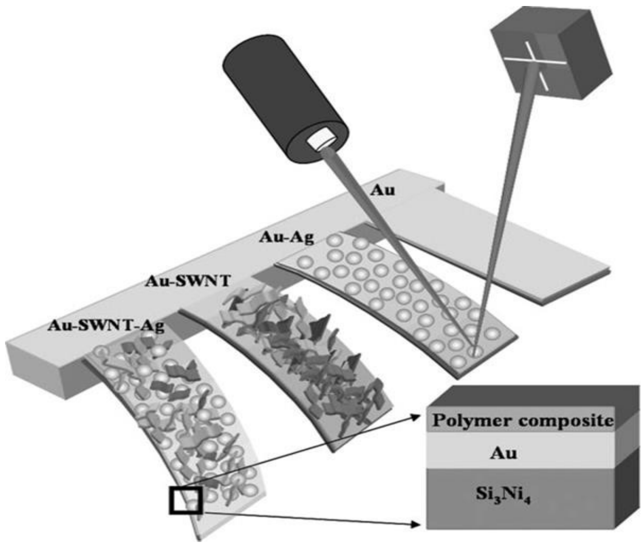

Fig. 7. Schematic of trilayered MCs coated with polymer nanocomposite for IR sens- ing application (*Source*: Lin et al., 2006).

materials with very different thermal-expansion coefficients. The thermally induced stresses resulted in reversible bending of the microcantilevers and optical readout of the cantilever deflection was measured. With this set-up the authors showed a four-fold increase in thermal sensitivity when compared to conventional metal–ceramic microcantilever.

Owing to the outstanding electronic and optical properties of single-walled carbon nanotubes (SWNTs), IR photoresponse of both individual SWNTs and SWNT films have been reported in a num- ber of studies (Freitag et al., 2003; Fujiwara et al., 2004; Itkis et al., 2006; Levitsky and Euler, 2003; Pradhan et al., 2008; Qiu et al., 2005). Pradhan et al. (2008) fabricated an IR-sensitive SWNT- polymer nanocomposite sensor. They embedded 5 wt.% SWNT into an electrically and thermally insulated matrix of polycarbon- ate. They concluded that for pure SWNT film, the thermal effect predominates and for SWNT-polycarbonate nanocomposites, the photo-effect predominates in the IR photoresponse.

2. Challenges

The field of polymer-based sensors is still nascent and there- fore, faces a dynamic set of challenges as the field evolves. Some of the major concerns of polymer materials include temperature and chemical stability, long term stability, and tolerance to high elec- tric field (Liu, 2007). Crosstalk is a primary issue in polymer-based sensor applications. For example, electroactive polymers such as ionic polymer–metal composites and other nanomaterial-based elastomers respond to changes in stress/strain, heat and humid- ity subsequently affecting their thermal, mechanical and chemical properties. Changes in material properties can complicate calibra- tion and adversely affect the sensor performance (Biddiss and Chau, 2006).

One of the major challenges in CP-based electrochemical sen-
sor design is to immobilize the transducer (CP matrix) onto an electrode substrate for effective signal transduction. In addition to the insufficient adhesion of CPs onto the substrate, conventional methods commonly used in the sensor fabrication: photolithog- raphy and e-beam lithography may not be compatible with CPs due to potential adverse affects of the techniques on the mate- rial properties of the polymer composites. Various other methods such as soft lithography, electrochemical deposition, and ink-jet printing have been proposed for patterning CP onto the electrode substrates. However, they have limited spatial resolution (Yoon and Jang, 2009). In order to overcome these limitations, studies have explored the possibility of chemical conjugating CP to the substrate surface (Dong et al., 2005a,b).

Failure mechanisms of conducting polymer films due to inter-
nal stresses have been extensively studied (Baumert et al., 2004; Benabdi and Roche, 1997; Campbell, 1969; Francis et al., 2002; Ohring, 1992; Wang et al., 2002; Wang and Feng, 2002). Some of the issues addressed in the literature are: effects of thickness and microstructures on the mechanical properties of CP films, failure behaviours on thermal and mechanical loading, and the heteroge- nous, highly localized stress and strain distributions, typical of flexible substrates,

found in elastomers. It is important to note that such failure models of CP-based films and coatings need to be care- fully examined in order to use them for commercial applications (Wang et al., 2009).

3. Conclusion

In summary, three different types of CPC- and ICP-based sensors relevant to clinical applications have been discussed in this paper. Owing to the flexibility, biocompatibility, and ability to deposit CPs onto desired geometrical surfaces and structures, CP-based sensing elements have immense potential to integrate into micro/nano scale devices for *in vivo* sensing and monitoring of bioanalytes. CP- based nanomaterials can be easily coupled with various chemical and/or biological species to obtain highly sensitive and selective responses. As discussed in Sections 2.1–2.3, we can further summa- rize that organic conducting polymers can be easily integrated into various microanalytical systems such as miniaturized microfluidic lab-on-chip devices or lab-on-tube devices (smart catheters) for as smart medical diagnostic tools and surgical aids. However, fabrica- tion protocols and challenges related to the processing and stability issues, as discussed in Section 3, can be improved by employ- ing suitable surface functionalization of nanomaterials, some of which were presented in Section 2.1. Moreover, advancements in the polymer-processing technologies and improvements in the material properties of CPs will eventually allow integration of CPs, semiconductors, and the underlying electronics collectively to be used as hybrid sensors for broad commercial applications in var- ious fields of biomedical engineering (prostheses, implants with feedback systems, biochips for personalized medicine, etc.).

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