Contents lists available at ScienceDirect

Biosensors and Bioelectronics

journal homepage: www.elsevier.com/locate/bios

Conductive polymer-based sensors for biomedical applications

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ARTICLE INFO

ABSTRACT

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. In this paper, we focus on CP-based sensor elements and the state-of-art of CP-based sensing devices that have potential applications as tools in clinical diagnosis and surgical interventions. Representative applications of CP-based sensors (electrochemical biosensor, tactile sensing 'skins', and thermal sensors) are briefly discussed. Finally, some of the key issues related to CP-based sensors are highlighted.

Contents

1. Introduction

Polymers form an integral part of our existence and everything that surrounds us—from the basic building blocks of life constituting of proteins, nucleic acids, and polysaccharides, to the commercial products obtained from automobile, construction and transportation industries, plastic toys and tools, reading glasses, etc. (Carraher, 2010). Most of these materials are composed of a combination of one or more materials to form polymer composites. Naturally occurring polymer composites such as bone (combination of ceramic calcium phosphate crystallites and collagen fibres can form either the dense, strong cortical bone or the spongy, shear-resistant cancellous bone), teeth (tooth enamel, dentin, and cementum, all of which contains varying percentages of inorganic hydroxyapatite crystals and organic material such as collagenous or non-collagenous proteins) and wood (a combination of cellulose and lignin), possess a unique combination of material properties and a broad spectrum of applications which the constituents alone cannot offer. In early twentieth century, plastic composites developed by mixing resin and filler material (wood flour, mica or glass) with enhanced strength and stability marked the beginning of synthetic polymer composites. Early investigations used conductive filler materials such as carbon black, graphite fibres, metal-coated glass fibre, metal particles or metal flakes for preparation of composite materials (Bhattacharya, 1986). Hence, a new class of materials called the conducting polymer composites (CPC)

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emerged opening up exciting new applications in several fields including bioelectronics. The CPCs typically consist of a combination of one or more non-conducting polymers and conductive-filler materials distributed throughout the polymer matrix. The conductivity 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 concentration, the filler particles no longer maintain physical contact with each other to provide continuous path for electron transport 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 conditions, 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 (oxidants) 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 conducting 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 electrical 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 conductivity 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 biosensors 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 material 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 surgeries, and the presence of biochemical components like glucose and cholesterol. CPs possess excellent electrical, chemical and mechanical 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, 2007; Schuhmann, 1995; Trojanowicz et al., 1997; Wen and Fang, 2008).

However, in this paper, we discuss the state-of-art of sensors based on two main classes of CPs—CPCs and ICPs. A brief history of CPC- and ICP-based sensors along with an overview of the basic principles behind sensing clinical parameters such as bioanalytes, force and temperature are presented in Sections 2.1–2.3 respectively. Section 3 highlights some of the challenges related to CP-based sensors. Finally, we summarize the key aspects of CP-based sensors and its potential use in the field of bioelectronics.

2. Representative applications

2.1. 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, antibodies, nucleic acids, hormones, drugs, viruses, neurotransmitters, pathogens and toxins. An electrochemical biosensor typically consists 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 biosensor) 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 transducers 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; Perĭez 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 materials 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 capability 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 mediators 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 immobilization) 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,
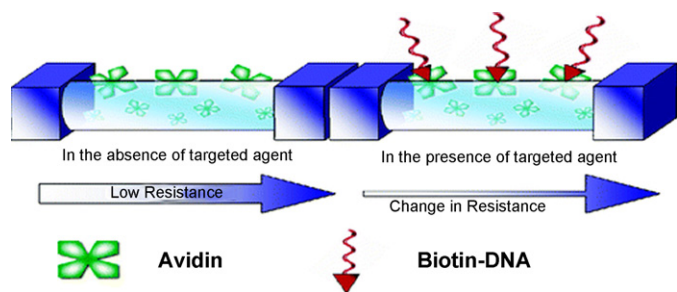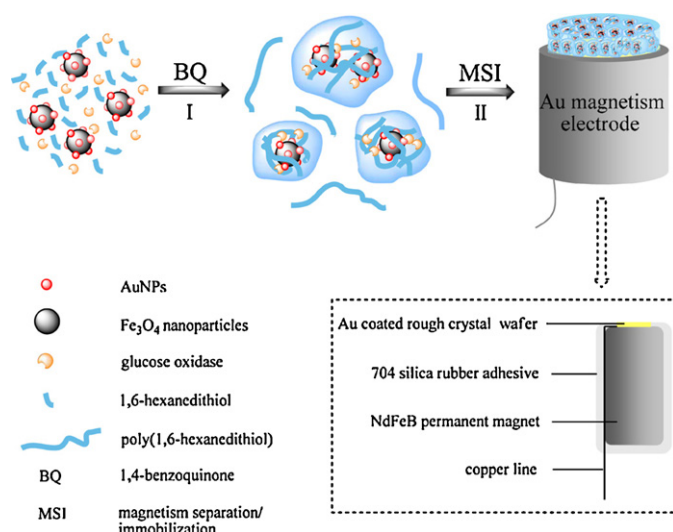
Fig. 1. Polymer (polypyrrole) nanowire on pre patterned electrodes and its application to biosensing was demonstrated (*Source*: Ramanathan et al., 2005).

Fig. 2. Schematic illustration of the construction of a Fe3O4–Au–PHDT–GOx modified Au magnetism electrode (*Source*: Zou et al., 2010).

the electropolymerized film remain stable in air, organic solvents and also, in aqueous solvents (Cosnier, 2007; Vidal et al., 2003). Electrodeposition also allows precise spatial control over localizing enzymes or other bioactive moieties into the polymeric matrix (Cosnier, 2005).

Initial research efforts on biosensors predominantly consists of enzymatic biosensors capable of detecting small molecules such as glucose, cholesterol, lactate, and urea. from millimolar to micromolar range. Another class of sensors known as affinity biosensors (immunosensors, DNA sensors, and receptor sensors) have also become increasingly popular due to their potential to sense biomolecules at extremely low concentrations, i.e. at the picomolar and even at the femtomolar level (Cosnier, 2005). Consequently, the immobilization of bioactive molecules in or on the transducer turns out to be a key aspect for obtaining highly efficient affinity biosensors (Geeta et al., 2006; Guimard et al., 2007). The deposition of biomolecules can be achieved by: non-covalent interaction (physical adsorption, mechanical entrapment and affinity bonding) and covalent linkages (chemical conjugation between functional groups of analyte and polymer). To date, both covalent and non-covalent modifications of the bio-recognition interface have been extensively explored and reviewed in detail (Bakker and Telting-Diaz, 2002; Cosnier, 2005; Gerard et al., 2002; Habermuller et al., 2000; Lange et al., 2008; Palmisano et al., 2000; Schuhmann, 2002). Some of the major concerns include sensor instability, poor loading capability and complicated manipulations (Zhang et al., 2007). In the past decade, magnetism-based immobilization of biomolecules (enzymes (Elyacoubi et al., 2006; Liu et al., 2005; Qu et al., 2007; Rossi et al., 2004; Yu et al., 2006; Zhang et al., 2007; Zou et al., 2010) and antibodies (Tang et al., 2006)) and cancer cells (leukaemia K562) (Jia et al., 2009) have been reported. A typical magnetism-based biosensor can be broadly divided into 3 units: (i) A biorecognition layer comprising of sensing elements (biomolecules or cells) attached to magnetic nanoparticles, and conducting filler materials (gold or silver nanoparticles, carbon powder, etc.) all entrapped within a polymer matrix, (ii) a permanent magnet onto which the biorecognition layer is mounted, and (iii) an insulator layer that packages all the units together. Magnetic immobilization is highly selective, efficient and convenient for biosensing studies. However, the biomolecules are often covalently immobilized over the surfaces of the magnetic nanoparticles and the loading is also limited. Zou et al. (2010) developed a magnetism-based biosensor with gold and magnetic nanoparticles entrapped in a polymer composite. They used a novel one-pot chemical oxidation synthesis and magnetism separation/immobilization (COSMSI) protocol for sensing glucose (Fig. 2). They reported higher enzyme-loading capability and a biosensing performance better than that of the biosensors based on conventional electropolymerization protocols (Zou et al., 2010).

Based on the mechanism used for signal detection and measurement, biosensors can be divided into several categories: amperometric, potentiometric, conductometric, optical, calorimetric, and piezoelectric. The most common types of transducers are amperometric (measures current) and potentiometric (measures potential). Amperometric biosensors measure the current produced by oxidation or reduction processes occurring in the bio-recognition layer (e.g., redox reaction of an enzyme substrate in an enzymatic biosensor). Investigations into amperometric biosensors based on polymeric nanotubes (Ekanayake et al., 2007) and nanofibers (Liu et al., 2007) have opened up new frontiers for electrochemical biosensors. For example, Ekanayake et al. (2007) developed PPy nanotube-array-based enzymatic biosensor for sensing glucose. PPy was deposited on a platinum plated nano-porous alumina substrate which enhanced adsorption of the enzyme-glucose oxidase and provided an increased surface area for the sensing reaction. Poly-2-amino-5-mercapto-1,3,4-thiadiazole (PAMT), has been recently reported as a very promising ICP for biosensing applications because of its outstanding electron-transfer properties. Kalimuthu and John (2009a,b) were the first to demonstrate the biosensing capability of PAMT film electrode-posited on glassy carbon substrate. The film was used for selective sensing of L-cysteine and folic acid. He et al. (2010) enhanced the amperometric sensing behaviour of PAMT film by using solid carbon paste as the substrate for electropolymerization of AMT which, in turn, was found to increase the amount of PAMT deposited on the substrate resulting in significant decrease in charge-transfer resistance during the sensing process. Potentiometric biosensors use ion-selective electrodes or gas sensing electrodes as physical transducers to measure electric potential due to concentration of analytes-of-interest. For example, urea is detected by ureases via the production of NH_3 which, in turn, interacts with PPy to produce an electric potential. A number of pH-sensitive polymers (poly (4,40-diaminodiphenyl ether), PANI, PPy, etc.) have been synthesized for potentiometric detection of analytes or ions-of-interest in solution (Trojanowicz, 2003). Electrochemical doping of CPs in biosensors enhances electron transfer by a combination of mechanisms such as redox reactions of the CPs, mobility of the dopant ions and the ion-exchange properties of the polymer. One of the factors contributing to an electric potential readout could be a product of a change in pH and the subsequent mobility of dopant ions within the polymer matrix triggered by an equilibrium process between the dopants and free ions in solution (Guimard et al., 2007). In summary, CP-based biosensors are likely to address the issues of biocompatibility for continuous monitoring of biological metabo-

lites and drug dosages, and the possibility of *in vivo* sensing (Rajesh et al., 2009).

2.2. 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 provides 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 minimally invasive surgery (MIS), which is the use of specifically designed surgical instruments and visual devices that allow surgeries 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 integration 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 feedback (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, capacitive 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 sensing elements in polymer skins (Beebe and Denton, 1994; Jiang et al., 1997; Wen and Fang, 2008), packaging of silicon-based sensing 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 sensors. 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;
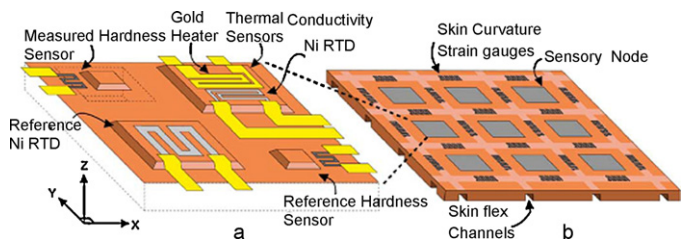

Fig. 3. A schematic of the multimodal tactile sensor. (a) A sensory node incorporates 4 distinct sensors: a reference temperature sensor, a thermal conductivity sensor, and contact force and hardness sensors. (b) Sensor nodes are arranged in an array to form skin. (*Source*: Engel et al., 2003a,b).

Yuji and Sonoda, 2006). De Rossi et al. (1993) reported a piezoelectric 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 polyhydroxybutirate (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 compliance of a contact tissue/object. The sensor set-up consisted of a pair of rigid and compliant cylinders, and two PVDF films. The compliance 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 application 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, relative hardness, thermal conductivity and temperature (Fig. 3). The sensing skin consists of an array of sensor nodes arranged on a flexible polyimide (Kapton) substrate. Each node comprised of 4 sensing elements: a thermal conductivity measurement unit, a temperature 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 sensitivity 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 sensor 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 polymer. 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 sensing 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
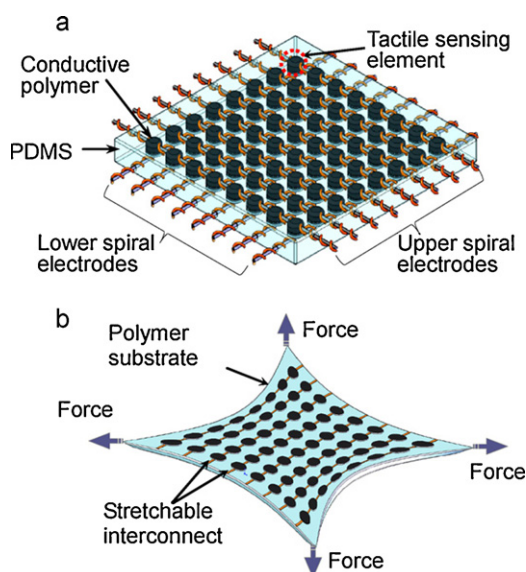
Fig. 4. (a) The schematic of the proposed tactile sensing array. (b) The proposed device under stretching. (*Source*: Cheng et al., 2010).

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 especially 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 conductive material that could be uniaxially and biaxially stretched by 70% without causing mechanical or electrical damage. The sensor 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 polymer was a composite material which is a blend of PDMS and a variety of conductive filler materials (copper, carbon black, cyclohexane 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.

2.3. Thermal sensors

Thermal readings have been used in medicine for several centuries. 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 circulating pyrogens produced by our immune system. With the advent of new technologies, thermal sensing has become a useful diagnostic 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 performances (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 thermal 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 functionalities of biological skin, temperature sensitive transducers used in electronic skins (*E*-skins) generally exploit two types of physical effects: thermoelectricity and pyroelectricity. Thermoelectric temperature sensors, also known as thermocouples, generate electricity 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 voltage 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 material 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 κ 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 composites 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 expansion 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 particles or ICPs) must form a conductive network within the insulating polymer matrix (Feng and Ellis, 2003). To achieve an optimal conductive 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 positive 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 polyimide 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 temperature resulting in a spontaneous polarization of the material. Temperature-dependent polarization slightly modifies the positions of the atoms within the crystal and produces a voltage across the material. This change in polarization with respect to tempera-
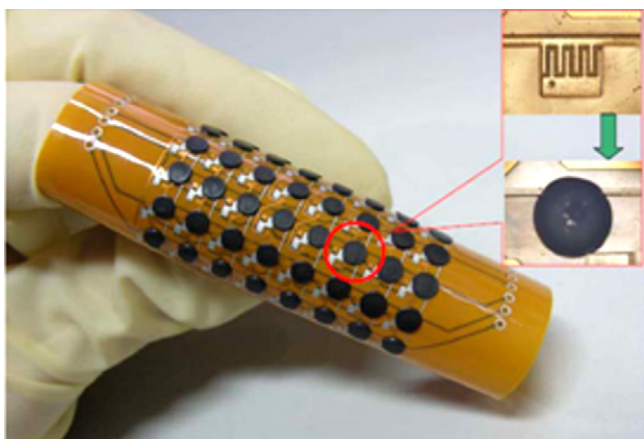
Fig. 5. Flexible temperature sensor array. The insets show the interdigitated electrode and composites on the electrode, respectively. (*Source*: Shih et al., 2010).

ture can be described as:

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

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 polymer, 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 coefficients of about 25 μC/m^2 K and 40 μC/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 materials 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 compared to metals and semiconductors. Most polymers absorb infrared radiation because of the vibrational resonance modes present in their organic bonds. Temperature-induced morphological 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 combinations 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.
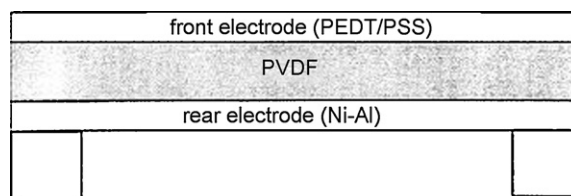
Fig. 6. A free standing, self-absorbing 28 μm PVDF pyroelectric sensor (*Source*: Setiadi et al., 1999a,b).

Several studies have investigated thermomechanical transduction 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 microcantilever. 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 fabricated bimorph microcantilever from novolak-resin photoresist and polysilicon for IR thermal sensing. Lin et al. (2006) developed a thermally sensitive microcantilever using a layer-by-layer structure of conductive polymer, metal and ceramic. The trilayered microbeam comprised of silicon nitride, a ceramic with low thermal expansion, an ultrathin film of gold followed by a topmost layer of chemically grafted polymer brushes (Fig. 7). The nanocomposite polymer structures of functionalized polyacrylonitrile (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
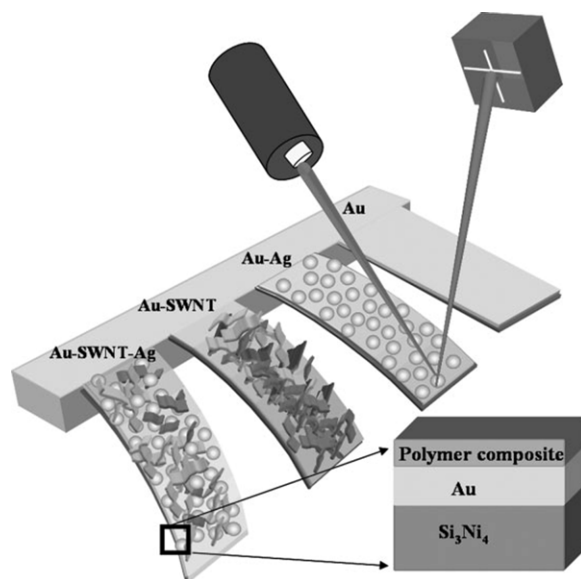
Fig. 7. Schematic of trilayered MCs coated with polymer nanocomposite for IR sensing application (*Source*: Lin et al., 2006).

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 number 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 polycarbonate. They concluded that for pure SWNT film, the thermal effect predominates and for SWNT-polycarbonate nanocomposites, the photo-effect predominates in the IR photo-response.

3. Challenges

The field of polymer-based sensors is still nascent and therefore, 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 electric 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 humidity subsequently affecting their thermal, mechanical and chemical properties. Changes in material properties can complicate calibration and adversely affect the sensor performance (Biddiss and Chau, 2006).

One of the major challenges in CP-based electrochemical sensor 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: photolithography and e-beam lithography may not be compatible with CPs due to potential adverse affects of the techniques on the material 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 internal 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 heterogenous, 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 carefully examined in order to use them for commercial applications (Wang et al., 2009).

4. 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 sens-

ing 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 summarize 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, fabrication protocols and challenges related to the processing and stability issues, as discussed in Section 3, can be improved by employing 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 various fields of biomedical engineering (prostheses, implants with feedback systems, biochips for personalized medicine, etc.).

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