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Chapter

Wearable Electromechanical Sensors and Its Applications

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Abstract

Wearable electromechanical sensor transforms mechanical stimulus into electrical signals. The main electromechanical sensors we focus on are strain and pressure sensors, which correspond to two main mechanical stimuli. According to their mechanisms, resistive and capacitive sensor attracts more attentions due to their simple structures, mechanisms, preparation method, and low cost. Various kinds of nanomaterials have been developed to fabricate them, including carbon nanomaterials, metallic, and conductive polymers. They have great potentials on health monitoring, human motion monitoring, speech recognition, and related human-machine interface applications. Here, we discuss their sensing mechanisms and fabrication methods and introduce recent progress on their performances and applications.

Keywords: wearable, electromechanical sensor, health monitoring, fabrication, mechanism

1. Introduction

With the rapid development of information technology, the Internet of Everything turns more critical in the next technological revolution. Wearable devices, which have the advantages of good portability, easy to carry, and multifunctional capability, are considered as the basic hardware in the future, which show great potential on many applications, including medicine, healthcare, robotic systems, prosthetics, visual realities, professional sports, as well as entertainment. In recent years, much efforts have been devoted to developing wearable sensing technologies. Various kinds of wearable sensors have been proposed and demonstrated in lab, from single functional sensors, such as temperature [1], pressure [2], strain [3], optical [4], and electrochemical sensors [5], to multifunctional sensors, such as tactile and electronic skin [6]. Among these wearable sensors, wearable electromechanical sensors including strain and pressure sensor have attracted more and more attentions due to its clear mechanism, low cost, low power consumption, and high performance [7]. Through integrating wearable strain and pressure sensor with other sensors, tactile sensor [8] and electronic skin [9] have been realized. High-performance wearable electromechanical sensor can monitor the tiny change of strain and pressure, which is useful in many fields.

Traditional electromechanical sensor is usually fabricated with brittle materials, such as silicon and metal. Though flexibility can be improved by structural design, their performance is still limited. Thus, many new materials have been developed. The materials used in wearable electromechanical sensor consist of sensing and supporting material. Most of the progresses are focusing on the development of

new sensing materials. Structural design is also an effective strategy to improve the performance. Fabrication method is also the significant aspect. Many traditional techniques are utilized, such as screen printing, contact printing, electrospinning, and spray coating [10]. Moreover, wearable electromechanical sensor has been successfully demonstrated on a lot of applications, such as health monitoring, disease diagnosis, behavior correction, alarm of accident falls, human-machine interfaces, and even speech recognition.

The present chapter will discuss their basic working mechanism, fabrication methods, and applications of wearable electromechanical sensors and challenges facing the progress.

2. Working mechanisms of a wearable electromechanical sensor

Firstly, we discuss the working mechanism of a wearable electromechanical sensor. Based on their working mechanisms, it can be classified into piezoresistive, capacitive, iontronic, and piezoelectric sensor, as seen in **Figure 1** [11].

2.1 Wearable piezoresistive sensor

Figure 1a shows the mechanism of piezoresistive sensor. It transfers mechanical stimuli into resistance signal. The factors resulting in resistance change depend on the property of materials utilized and their structures, including geometrical effect, structural effect, and disconnection mechanism.

2.1.1 Geometrical effect

Geometrical effect means that the resistance change is caused by geometrical change, which is mainly due to Poisson's ratio (υ). Poisson's ratio (υ) is a fundamental parameter of materials, meaning that materials tend to contract in transverse direction of stretching when they are stretched. The resistance of a conductor is represented by:
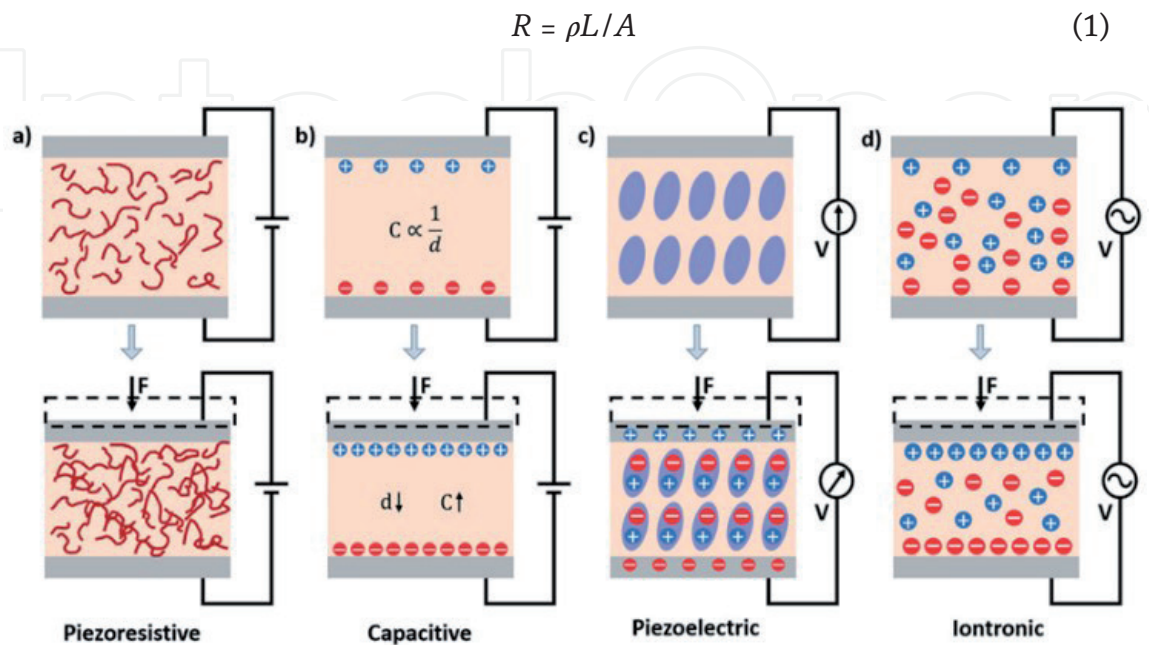
$$R = \rho L / A \tag{1}$$

Figure 1.
Schematics illustrating the different modalities of wearable electromechanical sensors. (a) Piezoresistivity, (b) capacitance, (c) piezoelectricity, and (d) iontronic.

where ρ is the electrical resistivity, L is the length, and A is the cross-sectional area of the conductor. When strain or pressure is applied, the length increases and cross-sectional area would be changed due to the shrinkage of materials, resulting in change of the resistance. Geometrical effect is usually limited compared to other factors.

2.1.2 Structural effect

Structural effect is defined as the change in the resistance caused by the structural deformations. This is usually observed in semiconducting materials. When strain or pressure is applied, the crystal structure especially interatomic space is changed, resulting in the change of the bandgap, which may increase the resistance of materials to few times [12]. For example, individual carbon nanotube (CNT) [13] shows ultrahigh resistivity change owning to their chirality and change in barrier height, respectively. However, compared with total resistance change, the part is usually low because strain applied on individual nanoflake is always small. In addition, the large elastic mismatch and weak interfacial adhesion strength between nanomaterials and polymers also make nanoflakes almost free from deformation.

2.1.3 Disconnection mechanism

The disconnection mechanism means that resistance change is caused by disconnection process between adjacent nanoflakes. It consists of three situations under different strains or pressures, which are contact area change, tunneling effect, and crack propagation.

When the applied strain or pressure is small, contact area changes between adjacent nanoflakes dominants. The electrons mainly pass through overlapped nanoflakes within the percolation conductive network. When the applied strain or pressure increases and fully pull some adjacent nanoflakes apart, the electrons still can pass through them because the distance between them is small enough. This phenomenon is called tunneling effect, and the distance is called tunneling distance. The tunneling resistance between two adjacent nanoflakes can be approximately estimated by Simmons's theory [14]:

$$R_{\text{tunnel}} = \frac{h^2 d}{Ae^2 \sqrt{2m\lambda}} \exp\left(\frac{4\pi d}{h}\sqrt{2m\lambda}\right) \qquad (2)$$

where A, e, h, d, m, λ represent the cross-sectional area of the tunneling junction, single electron charge, Plank's constant, the distance between adjacent nanoflakes, the mass of electron and the height of energy barrier for insulators, respectively. It can be found that the distance between adjacent nanoflakes dominates the tunneling resistance. When there is no electron pass through by tunneling, the distance is defined as cut-off tunneling distance. The cut-off distance is usually several nanometers. When the applied strain or pressure is large enough, crack is formed, leading to rapidly increasing of resistance. Strain or pressure leads opening and enlargement of cracks, critically limiting the electrical conduction due to the separation of several crack edges.

2.2 Wearable capacitive sensor

As **Figure 1b** shows wearable capacitive sensor is based on capacitance change of capacitor. Among different capacitors, the most popular architecture is the parallel-plate configuration because it is easy to be fabricated and its model is simple. The capacitive change can be expressed by the classic equation:

$$C = \kappa \frac{A}{d} \qquad (3)$$

in which κ, A, and d represent the permittivity of the medium between two plates, the overlap area, and the distance between two plates, respectively. When any of them is changed by the mechanical stimulus, the capacitance would be changed.

For capacitive strain sensor, when the strain ε is applied, the length of capacitor along the strain direction would be increased, which is expressed as $(1 + \varepsilon)l_0$, while the width and thickness of dielectric layer would be decreased, which is expressed as $(1 - \nu_{electrode})w_0$ and $(1 - \nu_{dielectric})d_0$, respectively. The $\nu_{electrode}$ and $\nu_{dielectric}$ are used to represent the Poisson's ratios of flexible electrodes and dielectric layer, respectively. If both flexible electrodes and dielectric layer have same Poisson's ratio, then the capacitance upon stretching could be calculated as:

$$C = (1 + \varepsilon)C_0 \qquad (4)$$

The equation indicates that the capacitance of capacitive strain sensor is linear with the applied strain. However, the linear relationship is only suitable for limited strain range. When the applied strain is higher than certain value, the relationship between different axes cannot be obtained simply by the Poisson's ratio.

For capacitive pressure sensor, the sensitivity (S) of capacitance to pressure is given by:

$$S = \delta(\Delta C/C_0)/\delta P \qquad (5)$$

where ΔC is the variation of capacitance (C–C$_0$) and P presents applied pressure. The most popular structure for the wearable pressure sensor is interlock structure, which is hard to make accurate analysis.

2.3 Iontronic sensors

As **Figure 1c** shows, iontronic sensor is based on the iontronic interface sensing mechanism. The iontronic interface usually exists at the nanoscale interface between the electrode and the electrolyte. The electrode forms ionic-electronic contact with ionic gel. The electrons on the electrode and the counter ions from the iontronic film accumulate and attract to each other at a nanoscopic distance, leading to an ultrahigh unit-area capacitance. Compared to traditional parallel plate capacitive sensors, iontronic sensor has a higher surface area and its electrical capacitance is at last 1000 times larger. This excellent property is suitable for wearable electromechanical sensors. In addition, this special mechanism enables iontronic sensor immunity to environmental or body capacitive noises. So far, ion gels and ionic liquids are the most popular materials for iontronic sensor.

2.4 Piezoelectric sensors

As **Figure 1d** shows, the sensing mechanism of piezoelectric sensor is piezoelectric effect. Piezoelectric means that electric change accumulates in piezoelectric materials when mechanical stress is applied. Many materials have piezoelectric property, such as crystals, certain ceramics, and even biological matter. When strain or pressure is applied, there is a change in electrical polarization inside the material, resulting in a change in surface charge (voltage) at the surface of the piezoelectric material. In general, the electrical signal of piezoelectric sensor is voltage, which can be collected by measuring two different surfaces.

3. Performance of wearable electromechanical sensor

3.1 Basic parameters of wearable electromechanical sensor

3.1.1 Sensitivity and linearity

Sensitivity is the magnitude of electrical response to measured mechanical stimulus, which is an important parameter. For strain sensor, sensitivity is called gauge factor (GF), which is defined as $GF = \Delta R/R_0$ for resistive type and $GF = \Delta C/C_0$ for capacitive type. For pressure sensor, pressure sensitivity (PS) is defined as $PS = (\Delta R/R_0)/P$. Sensitivity can be affected by functional material, sensing mechanism, and structural configuration. The materials with large piezoresistive or piezoelectric coefficient are desired. Tunneling effect and crack/gap structures in piezoresistive sensors have been proven to be effective in promoting sensitivity. However, most highly sensitive sensors always show limited stretchability.

Linearity characterizes degree of deviation from linear relationship between electrical signals and mechanical stimulus. High linearity is convenient for the calibration and data processing process. However, there is always a contradiction between sensitivity and linearity because crack propagation and tunneling-effect-induced resistance change are usually exponential. For instance, piezoresistive strain sensors often exhibit varied sensitivity in different strain ranges, which is induced by the nonlinear heterogeneous deformation. In addition, capacitive sensors with microstructured dielectric also suffer the similar problem.

3.1.2 Hysteresis and response time

Hysteresis and response time are another two important parameters in evaluating dynamical performance of electromechanical sensor. Hysteresis means the dependence of the performance on its history, which should be reduced or avoided. In general, capacitive sensors show immediate responding to the variation of overlapped area, featuring a lower hysteresis. Meanwhile, piezoresistive sensors have slower response due to the interactive motion between sensing material and polymer substrate. The interfacial binding between sensing material and substrate greatly affects the optimization of hysteresis. The full recovery of sensing material position is hindered by the interfacial slide, leading to a high hysteresis behavior. Meanwhile, to avoid the friction-induced buckling and facture in sensing materials, a weak adhesion is needed. It is reported that using low viscoelastic polymer substrate and improved configuration can partially eliminate hysteresis. However, it is still a large challenge to optimize hysteresis by novel material and structural engineering. Response time illustrates the speed to achieve steady response to applied mechanical stimulus, and response delay exists in nearly all composite-based sensors because of the viscoelastic property of polymers. Relatively, piezoresistive device has a larger response time than others because it needs more time to reestablish percolation network in resistive composites. In addition, lower modulus materials are popular for wearable electromechanical sensor, which can further decrease the response speed of resistive sensors. Moreover, based on structural design, the newly developed crack-based piezoresistive sensors show an appealing response time (about 20 ms) because cracks can reversibly connect and disconnect with loading and unloading of mechanical stimuli [15].

3.1.3 Durability

Durability is the ability to remain its performance, without requiring excessive maintenance or repair, when it is normally used. It is usually measured by cyclic stability for wearable electromechanical sensor. Cyclic stability is sensor endurance to periodic loading and unloading cycles. The sensing material film on polymer substrate is easy to form buckling, facture, and even stripping after enough cycles, which results in cyclic instable problem. For example, the sensitivity of graphene woven fabric (GWF) strain sensor decreases 24% after about 1000 cycles from 0 to 2% [16].

Endowing sensor with self-healing is a novel way to promoting durability. Several works have been reported on wearable electromechanical sensor. **Figure 2a** shows a stretchable self-healing piezoresistive strain sensor using single wall carbon nanotube (SWCNT) in self-healing hydrogel (SWCNT/hydrogel) as the conductive sensing channel [17]. The cutting groove is partially healed after 30 s and totally restored to normal after 60 s at room temperature without any external assistance. It also shows the repetitive cutting-healing processes with five cycles at the same location. The average efficiencies are 98 ± 0.8% for the five self-healing cycles within about 3.2 s, indicating that the SWCNT/hydrogel possesses significant and repeatable electrical restoration performance. **Figure 2b** shows that a self-healing sensor with tunable positive/negative piezoresistivity is designed by the construction of hierarchical structure connected through supramolecular metal-ligand coordination bonds [18]. The electrical resistance of the repaired samples only slightly increases after multiple cutting/healing cycles. However, the increase of electrical resistance is neglectable, which is lower than one order of magnitude, indicating its excellent electrical self-healing ability. The high-healing efficiency is estimated to be 88.6% after the third healing process, and the healed wearable strain sensor still show good flexibility, high sensitivity, and accurate detection capability, even after bending over 10,000 cycles.

3.1.4 Biocompatibility

Wearable electromechanical sensors are usually directly used on human skins, so biocompatibility is also important. The main danger comes from sensing materials, which is usually nanomaterial other than substrate materials, which is a polymer. For example, it has been reported that injecting large quantities of CNTs into mice lungs could cause asbestos-like pathogenicity because of the small size and needle-like morphology of CNT [19]. To improve the biocompatibility, organic active materials, such as polypyrrole (PPy) and poly(3,4-ethylenedioxythiophene)
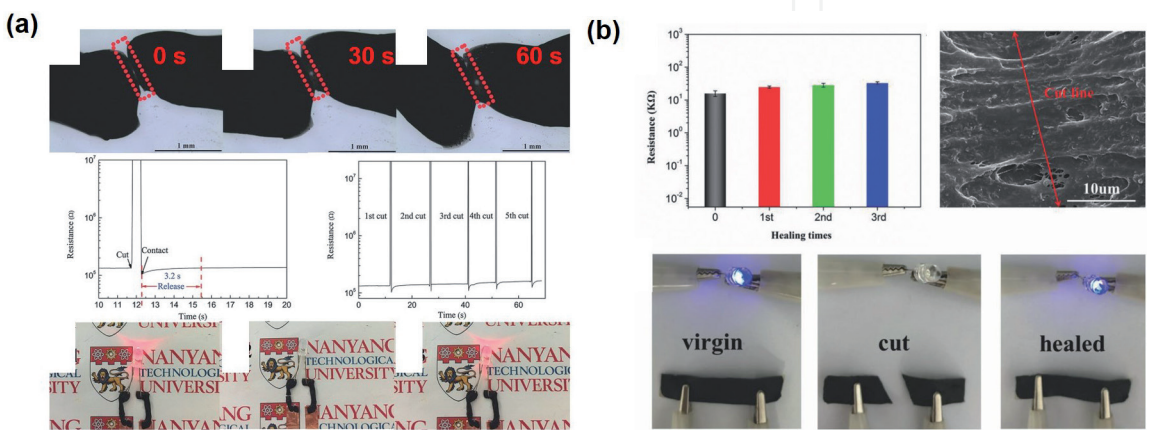

Figure 2.
(a) Self-healing properties of SWCNT/hydrogel-based strain sensor. (b) Electrical self-healing properties of supramolecular-elastomer-based strain sensor.

(PEDOT), have generally been used. The carbonized cotton or silk also presents great potentials in constructing biocompatible wearable sensors [20].

3.1.5 Self-power

Power is the basic element for wearable system. Wearable devices with self-power ability attract more and more attentions, which can greatly extend their application scenarios and is particularly suitable for long-lasting wearables. Self-power wearable electromechanical sensor has been demonstrated so far using triboelectric [21], photo-voltaic [22], piezoelectric [23], radiofrequency, thermoelectric (TE) systems [24], and others [25]. Among them, TE technology is rather attractive because of the utilization of conjugated polymers as the active component, which is also flexible, enabling a new generation of novel, low-cost, low-powered wearable electromechanical sensors [26].

3.2 Materials for wearable electromechanical sensor

3.2.1 Materials for substrate

Substrate is mainly responsible for flexibility and stretchability, and directly determines the comfort level and long-term reliability. Polydimethylsiloxane (PDMS), a commercial silicone elastomer with intrinsic high stretchability (up to 1000%), nontoxic, nonflammability, hydrophobicity, and good processability, has been frequently used. Though cannot be stretched for its relatively high modulus (about 2~4 GPa), polyethylene terephthalate (PET) features good transparency (>85%), high creep resistance, and excellent printability. Silicone elastomers including Ecoflex, Sylgard, Dragon Skin, and Silbione are biocompatible and their maximum stretchability is up to 900%. They are suitable flexible substrate because of their strong adhesion onto target surfaces. Ecoflex® rubber is a newly developed, highly stretchable and skin safe silicone with better stretchability and lower modulus, which has been used in the sensors requiring more severe flexibility and stretchability. Polyimide (PI) is another frequently used substrate because it can maintain flexibility, creep resistance and tensile strength under the condi-tion of high temperature (up to 360°C) and acids/alkalis. Thus, PI is compatible with micromanufacturing process and many types of wearable electromechanical sensor are possible to be designed and implemented on it. Natural materials are also explored and developed to produce flexible substrate because they are easily biodegraded, such as cellulose paper. Moreover, the natural textiles, like silk and cotton, are also highly desirable substrate materials [41].

3.2.2 Materials for active elements

3.2.2.1 Carbon nanomaterials

Carbon nanomaterials including graphite, CNT and graphene, have been widely used in fabricating wearable electromechanical sensors. Graphite is a conductor and attracts more and more attentions with development of pencil-on-paper electronics [54]. Graphite flakes in pencil lead is easy to be deposited on paper surface by the physical friction between lead tip and porous cellulose paper. Moreover, structural edges in graphite flakes results in a strain-induced resistance variation of pencil traces, making them suitable for strain sensor. The contact area between graphite flakes increases by compressing the trace and decreases when the tension strain is applied, leading to the decrease or increase of resistance. The wearable strain sensor fabricated with pencil-on-paper shows high GF up to 536.61 [27].

CNT are allotropes of carbon with a cylindrical nanostructure, which possesses excellent electrical conductivity and mechanical properties. It has been demonstrated that a single CNT shows strong structural effect and has a GF higher than 1000. However, wearable electromechanical sensor fabricated with single CNT is difficult and hard to realize mass production. Thus, CNT is usually intermingled into polymer substrates and its excellent conductivity plays an important role in electromechanical sensor construction. Wearable capacitive and piezoresistive electromechanical sensors have all been demonstrated by depositing CNT onto substrate or forming composite with polymers. For the piezoresistive composite sensor, the resistance change is mainly due to the strain-varied intertube tunneling resistance. The maximum GF can be achieved when the concentration of CNT is near the percolation threshold (PH). When the CNT loading is much lower than PH, the distance between adjacent CNTs is larger than their cut-off distance and there is almost no tunneling resistance. On the contrary, when CNT loading is much higher than PH, the CNTs can form dense 3D network and most of CNTs would connect with each other, resulting in a small intertube resistance. In this case, the contact resistance dominates the behavior, which will significantly decrease the GF. For piezoresistive film sensor, the variation of resistance gains almost a tenfold increment compared with nanocomposite type, but its cycle durability is not favorable enough because of unexpected cracks and desquamations. CNTs are also used to form wrinkle structure on a soft substrate via heating of the film or a prestrained substrate and are utilized to fabricate high-performance wearable strain sensor.

Due to outstanding electroconductibility, excellent mechanical properties, great thermal characteristic and optical transmittance, graphene becomes the most promising sensing material for the development of wearable electromechanical sensor [28]. Graphene has been developed as electrode material for capacitive sensor and filler for piezoresistive sensor. A variety of graphene electromechanical sensors with different forms have been demonstrated, including porous foams, flakes, ripples, woven fabrics, and films. For example, the GWF film, which can be fabricated either by CVD or dip coating, consisted of many overlapping microribbons and features a good trade-off between sensitivity and stretchability, making it suitable for wearable strain sensors. It shows fascinating stretchability (a tolerable strain up to 57%) and sensitivity (GF = 416 for $0 < \varepsilon < 40\%$, and GF = 3667 for $48 < \varepsilon < 57\%$) by encapsulating the obtained GWFs in natural rubber latex [29].

3.2.2.2 Metal materials

Metal possesses excellent electrical conductivity and has been widely used in wearable electromechanical sensors. There are four forms of metal developed, which are nanowires, nanoparticles, stretchable configurations, and liquid state at room temperature. Nanowires (NWs) and nanoparticles (NPs) are usually used to prepare piezoresistive composites or conductive ink. For example, silver nanowire (AgNW) can be embedded into PDMS to build resistive-type strain sensor. Because the adhesion between AgNWs and polymers is not as strong as carbon nanomaterials, AgNW interconnection is easy to be broken. The resistance will irreversibly increase after buckling and wrinkling if the AgNW film is just simply coated on the surface of polymer. In addition, AgNWs are easy to be oxidized. Therefore, AgNW layer is often sandwiched between two polymer layers, ensuring AgNWs to move back along their determined paths and be free from oxidation [23]. The stretchable configurations of metal are on the basis of the strategy "structures that are flexible and stretch." Coiled buckled, serpentine and woven structures have been utilized to endow flexibility and stretchability to metals. The liquid metal, like Ga and its alloys, maintains the liquid state at room temperature. With the help of microfluidic

techniques, liquid metals show a great potential on wearable sensors. When strain or pressure is applied, the microchannel geometry will be changed, leading to a significant variation in the sectional area and length of liquid metal resistor. The change of electric resistance can reach as much as 50%.

3.2.2.3 Polymer

Conductive polymers possess favorable electroproperties and can participate in building sensing materials. An attractive feature of conductive polymer is the mechanical similarity between them and many insulated substrate polymers. PEDOT-based polymers are the most common sensing materials for their thermal stability, high transparency, and tunable conductivity. Among them, poly(3,4-ethylenedioxythiophene)-polystyrene sulfonate (PEDOT:PSS) is one of the promising conductive polymers due to its excellent solubility in water. However, the dried PEDOT:PSS film is easy to form hard particles inside, which may induce fissure and then decrease electrical conductivity. It is not suitable for continuous bending and stretching. To solve this problem, porous substrates have been developed for printing and permeating PEDOT:PSS ink, such as fabrics and cellulose paper, which can greatly promote their adhesion. This strategy greatly improves the stability of wearable electromechanical sensor fabricated with PEDOT:PSS ink [30]. The polyvinyledenedifluoride (PVDF) is another appealing sensing material with many attractive properties, such as piezoelectric property, especially appropriate for piezoelectric wearable electromechanical sensors. Moreover, other conductive polymers such as PPy, poly(3-hexylthiophene-2,5-diyl) (P3HT) and PANI have also been utilized to fabricate wearable sensors [31]. More recently, ionic liquid (IL), a kind of salt that keep liquid state at room temperature, has attracted extensive attention [32]. Similar to liquid metals, IL can also be embedded in PDMS-based microchannels to fabricate wearable electromechanical sensor.

3.3 Performance of wearable electromechanical sensor

3.3.1 Wearable strain sensor

Wearable strain sensor converts strain into electrical signal. Many applications, such as human health monitoring, require enough stretchability range from tiny deformation (small than 1%) to large deformations (as large as 100%) and high sensitivity. There are two main strategies to enhance the sensitivity. One is choosing proper sensing materials. Various kinds of nanomaterials are tested, as seen in **Table 1**. For example, by coating graphene on woven fabric structure, a maximum elongation of 57% and a GF of 416 and 3667 at lower and higher strains are achieved. Combining graphene and nanocellulose into nanocomposite, it shows ultrahigh sensitivity with GF of 502 at 1% strain and 2427 at 6% strain.

The second strategy is structure engineering. As discussed in above section, cracks can greatly enhance the change of resistance. Network cracks formed in multilayer CNT films on PDMS composite result in both high gauge factor (maximum value of 87) and a wide sensing range (up to 100%) of the strain sensor, which allows the detection of strain as low as 0.007% with excellent stability (1500 cycles) [27].

To improve stretchability, many strategies have been developed. One strategy is using intrinsically flexible materials and the relative stiff components bridged with highly flexible interconnects [48]. When the intrinsic stretchability of flexible material is not enough, structural engineering can be used to further enhance their stretchability. The fragmented structure with connected islands can form a lot of cracks, which can relieve most of the applied strain through opening and enlargement

Material	Type	Sensitivity	Stretchability	Linearity	Durability (cycles)	Refs
AgNW	Strain	150,000	60%	0.989	200	[33]
AgNW	Pressure	1.54 kPa^{-1}	0.6 Pa-115 kPa	linear	5000	[34]
AuNW (gold nanowire)	Strain	70	250%	Nonlinear	500	[35]
AuNW	Pressure	1.14 kPa^{-1}	13 Pa-5 kPa	Linear	5000	[36]
Carbon black	Strain	647	20%	Nonlinear	200	[37]
Carbon black	Pressure	4.2 kPa^{-1}	0–30 kPa	0.996	30,000	[38]
Carbon nanofiber	Strain	72	300%	Nonlinear	8000	[39]
Carbon nanofiber	Pressure	4.2 kPa^{-1}	1.0 Pa-2 kPa	Nonlinear	10,000	[40]
Carbonized silk	Strain	9.6 (0–250%) 37.5 (250–500%)	500%	Nonlinear	10,000	[41]
Carbon nanotube	Strain	80	100%	Nonlinear	1500	[42]
Carbon nanotube	Pressure	0.209 kPa^{-1}	5.0 Pa-50 kPa	Nonlinear	5000	[43]
Graphene	Pressure	1.2 kPa^{-1}	0–25 kPa	Linear	1000	[44]
Graphene	Strain	1054	26%	Nonlinear	500	[45]
Mxene	Strain	64.6 (0–30%) 772.6 (30–70%)	130%	Nonlinear	5000	[46]
Mxene	Pressure	4.05 kPa^{-1} (0–1.0 kPa) 22.56 kPa^{-1} (1–3.5 kPa)	0–3.5 kPa	Nonlinear	10,000	[47]

Table 1.
Performance of wearable electromechanical sensor fabricated with typical nanomaterials.

of cracks. Deformable structures are widely used. For instance, the horseshoe and filamentary serpentine have been patterned with nanomaterials, which can accommodate large strain [49, 50]. Porous structures such as sponge and foam are also employed to improve the stretchability [51]. Wrinkled structure based on CNT film is produced and integrated on an Ecoflex substrate, allowing conductivity up to 750% elongation, an approximate 60 times increase versus nonwrinkled films [52].

Significant progress has been achieved on the sensitivity and stretchability, but there are some challenges still existing. Most resistive wearable strain sensors suffer from at least one of these problems, which are nonlinear response, large hysteresis, and irreversibility. The irreversibility mainly origins from partial slides back of sensing materials and irreversibly recover of cracks. Hysteresis is mainly caused by the visco-elasticity of polymers and the friction between the sensing materials and the polymer matrix. The rearrangement of sensing materials and opening of cracks are also responsible for time delay between electrical output and mechanical input. Nonlinear response mainly results from crack propagation and tunneling effect, which is always exponential as discussed above. Therefore, the performance of resistive wearable strain sensor should be evaluated from more aspects in further research.

Compared with resistive wearable strain sensor, capacitive strain sensors possess good linearity with low hysteresis, fast response, and are less susceptible to overshoot and creep. Nanomaterial-based stretchable conductors are usually used as the electrodes for capacitive strain sensors. Highly stretchable silicone, such as PDMS, Dragon Skin, and Ecoflex are commonly used as the dielectric layer sandwiched between two electrodes. For example, a capacitive strain sensor is fabricated with stretchable AgNW/PDMS conductors as the top and bottom electrodes and Ecoflex as the dielectric material [53]. The GF of this sensor reaches 0.7 and its stretchability is up to 50%. Moreover, it also has a good linearity. While capacitive strain sensors exhibit smaller GFs than the resistive strain sensors, they are ideal for applications where the strain is relatively large. In addition, the GFs of capacitive strain sensors remain constant in the entire strain range.

3.3.2 Wearable pressure sensor

Wearable pressure sensor converts pressure into electrical signal. Pressure sensor can be fabricated with interlocked structures, percolative networks of nanomaterials, microfabricated structures (e.g., micropyramids, micropillars), porous structures (e.g., sponges, foams, porous rubbers), and so forth. For example, **Figure 3a** presents a pressure sensor fabricated with interlocked microdome array. The contact between microdome increases when pressure is applied, thus decreasing the tunneling resistance [54].

To improve the sensitivity of piezoresistive pressure sensor, structural surface modification of the electrodes is an effective strategy. Incorporation of nano/microscaled structures can provide large changes in contact resistance, allowing for detections of smaller pressures. For example, through coating polyurethane sponge with graphene to form fracture structure, a two-order of magnitude increase in sensitivity within the 0–2 kPa regime is demonstrated compared with no fracture one [55].

For the capacitive pressure sensor, the separation between two electrodes decreases with the pressure, resulting in an increase in capacitance. The property of dielectric materials almost determines the pressure sensitivity. Lower elastic modulus means a larger strain ε under a given pressure. The dielectric constant increased with pressure and low Poisson's ratio would all benefit the performance. High sensitivity of 0.8 kPa^{-1} has been reported by using a GO-based low elastic modulus foam as the dielectric material [56]. There are several methods been demonstrated to fabricate highly deformable dielectric materials, including using commercial
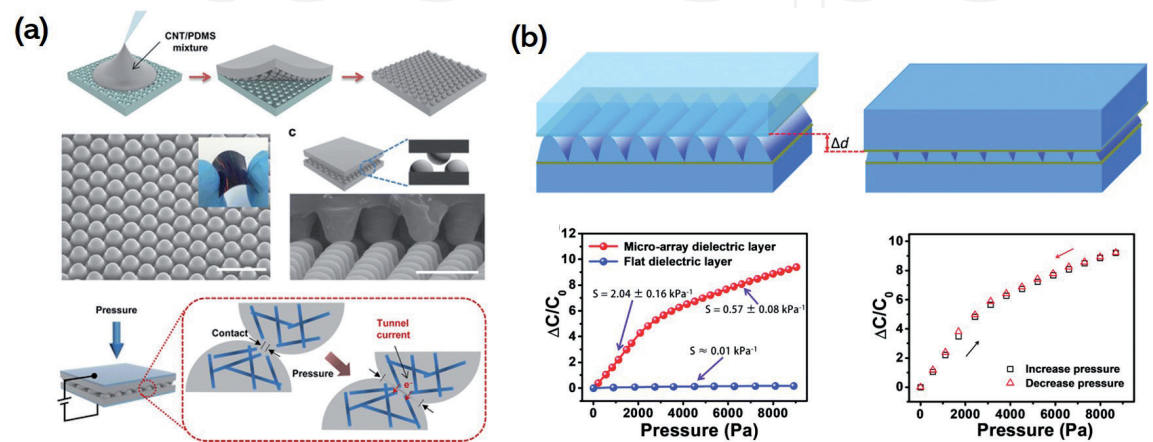

Figure 3.
(a) Schematic of the fabrication procedure and mechanism of pressure with interlocked microdome arrays.
(b) Response characteristics of the flexible capacitive pressure sensor based on the PDMS microarray dielectric layer.

porous tapes, using special molds (e.g., the surface of matte glass, a micromachined Si mold, or the surface of lotus leaf) to create microstructures in elastomers, using sugar cubes as the template to create porous elastomers and fabricating buckled structures through prestretching and releasing. As the dielectric constant of air is smaller than that of the dielectric material used for the sensor, the effective dielectric constant is increased under pressure when the air gap is compressed. For example, **Figure 3b** shows a flexible pressure sensor with high sensitivity been built, which is a typical sandwich structure by combining a microarrayed PDMS dielectric layer with PDMS substrates. The top/bottom electrode material is PDMS substrate coated with AgNWs, and the dielectric layer is a PDMS with microarray structure, which is used to improve the pressure sensitivity. The results show that it possesses high sensitivity (2.04 kPa^{-1}) in low-pressure ranges (0–2000 Pa), low detection limits (<7 Pa), and fast response times (<100 ms). Meanwhile, it also has excellent bending and cycling stability [57].

Progress has also been made on wearable piezoelectric and triboelectric pressure sensors. For example, it has been reported that a novel piezoelectric pressure sensor was fabricated through sandwiching freestanding electrospun polyvinyledenedifluoride-trifluoroethylene (PVDF-TrFE) nanofiber arrays [58] or electrospun PVDF-TrFE nanofiber between two electrodes. It can detect very tiny pressures as low as 0.1 Pa and has high sensitivity up to 1.1 V kPa^{-1} for pressure range from 0.4–2 kPa. In a representative work, a pressure-responsive triboelectric nanogenerator is used to gate the graphene transistors. Such graphene tribotronics showed a pressure sensitivity of ≈2% kPa^{-1} at a pressure of 10 kPa.

4. Fabrication technology of wearable electromechanical sensor

The wearable electromechanical sensor usually consists of three basic components, which are substrate, active elements, and electrode/interconnect. They are usually fabricated with different materials. During the fabrication process, combining the substrate and active elements is the key step. Basically, there are two situations. One is that sensing material forms uniform composite with polymer substrate, the other is that sensing material is attached on substrate and a clear interface exists. In this part, we will focus on the combination strategies for substrates and sensing elements, and some key processes for performance enhancement are also concerned.

4.1 Fabrication of wearable composite electromechanical sensor

For the composite electromechanical sensor, the substrate and sensing materials should be fabricated into composite. The key process is how to mix them and prepare uniform composite. The sensing materials are usually mixed with polymers by magnetically or ultrasonically stirring, and then the dried elastic composites can be prepared in bulk or film forms. The mixed composites have complex electromechanical features that are induced by the diversity of sensing materials and polymer and significantly depend on concentration of sensing materials and its distribution state. For example, the big electrical property of carbon black-silicone composite is mainly determined by carbon black concentration. The electrical resistance clearly increases with the applied uniaxial pressure when the concentration is about 0.08–0.09 wt%. By further increasing the concentration from 0.1 to 0.13 wt%, the change tendency of electrical resistance switches from increase to decrease. Finally, the electrical resistance starts to decrease with the uniaxial pressure with the concentration larger than 0.14 wt % [59].

4.2 Fabrication of wearable layered electromechanical sensor

For the wearable layer electromechanical sensor, the substrate and sensing materials are assembled into film layer by layer. Many techniques have been developed to assemble active material on substrate, including printing, coating, casting, and other methods.

Printing can simultaneously deposit and pattern many materials on various substrates without the need for sophisticated equipment and clean room. The wearable sensors can be printed with/without the help of masks, according to the specific implementation approach, as seen in **Figure 4a** [60]. The electrode pattern can directly be obtained by inkjet printing. Inkjet printing is an accurate, fast, and reproducible film preparation technique. Functional ink droplets are propelled onto different substrates by a nozzle. The functional inks should have proper solubility, viscosity, and surface tension. As a typical printing method, screen printing requires the help of mask and proper functional ink. During the process, screen openings are fully covered with functional by using fill blade or squeegee, and then it is transferred onto substrate surface. Finally, the mask is removed, and a patterned film is formed on the substrate by functional ink. This technique has been widely used in manufacturing sensing materials in electromechanical sensors.

Lithography is a pattern transferring method to realize diverse and ingenious geometries. This process firstly deposits functional layer onto the substrate and then etches the undesired areas by reagent solutions with the help of photolithography. Since photolithography and wet etching has high accuracy, the devices with sophisticated geometries and rich functionality can be obtained. Coating technique is another popular method because of its low cost and simplicity. There are different advantages for different coating methods. Dip coating can be used to any kinds of substrate and can control the thickness by dipping time. Spin coating is easy to form uniform film and can control the thickness by time and spin speed. Compared with spin and dip coating, spray coating can fully utilize the functional inks. **Figure 4b** shows a buckled sheath-core fiber-based ultrastretchable sensor fabricated with spray coating methods. The fiber wearable strain sensor possesses excellent stretchability higher than 1135% and fast response time (≈16 ms). Moreover, the performance is very repeatable and stable even after 20,000 cycles with loading/unloading test [47].

Novel techniques have been developed, such as laser scribed (LS) technique. Graphene oxide (GO) can be simultaneously reduced and patterned by laser [61]. Carbonating substrate material by one-step direct laser writing (DLW) has also been validated. Glassy and porous carbon structures have been produced from PI film via DLW. The DLW-based graphene possesses favorable electroconductibility, porousness, and superhydrophilic wettability. Directly drawing electronics with various instruments has recently become an alternative technique. This technique endows end-users the capability to design and realize sensors according to the "on-site, real-time" demands [62]. "Penciling it on" has been proved to be a simple, rapid, and solvent-free method for producing electronics [63]. Chinese brush pen is a possible more appealing writing instrument for sensor fabrication. Similarly, the animal hair bundle is first soaked into low-viscosity ink, and then the ink is uniformly coated on the substrate by well-controlled handwriting manner. Benefiting from excellent liquid manipulation of Chinese brush pen, sensing materials can be coated on different substrates without considering its rigidness and surface roughness. For example, a high-performance tattoo-like strain sensor has been fabricated with AuNWs/PANI ink writing by Chinese brush pen [64]. Various types of functional inks can be loaded in their reservoirs, including metal inks, liquid metals, and even organic mixtures. Sophisticated structures can be generated with controllable geometries on many substrates by using these two methods [65]. Wet spinning is
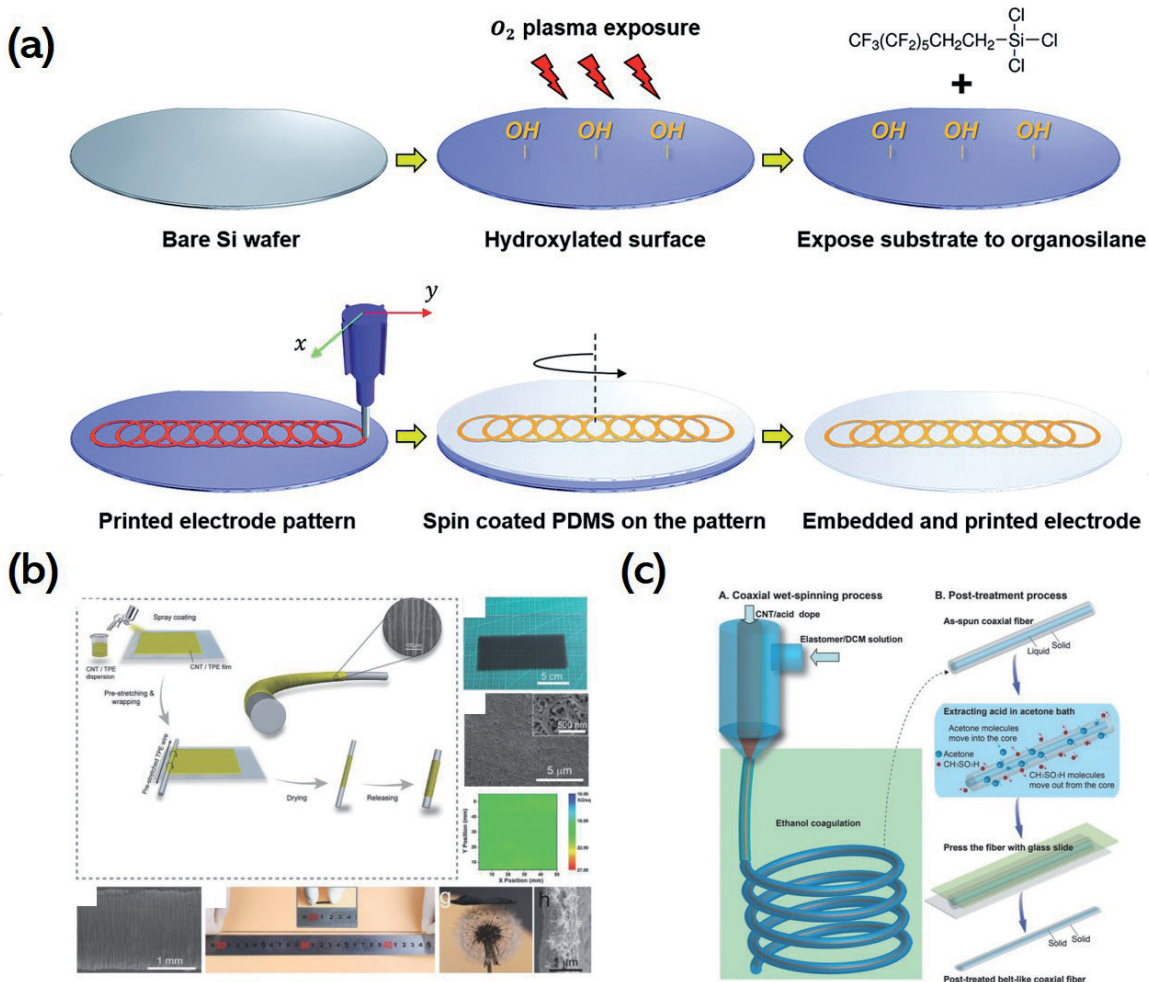
Figure 4.
Wearable electromechanical sensor fabricated with different techniques: (a) inkjet printing, (b) dropping casting, and (c) spray coating.

another special method to fabricate fiber shape wearable electromechanical sensor. **Figure 5c** shows a fiber strain sensor fabricated with coaxial wet-spinning and post-treatment process. The spinning nozzle has the coaxial inner and outer channels, respectively. The inner spinning dope is $SWCNT/CH_3SO_3H$, and the outer spinning solution is the solution of thermoplastic elastomer (TPE) in CH_2Cl_2. The SWCNT/ CH_3SO_3H dope from the inner channel and the TPE/CH_2Cl_2 solution from the outer channel are introduced into the ethanol coagulation bath simultaneously. A single TPE-wrapped SWCNT coaxial fiber is then wetspun and collected successfully. The sensors attain high sensitivity (with a gauge factor of 425 at 100% strain), high stretchability, and high linearity.

4.3 Fabrication of wearable 3D electromechanical sensor

For the wearable 3D electromechanical sensor, the substrate and sensing materials are combined into 3D structure. The first method introduced is microscale modeling. It is often utilized to fabricate different microstructures in substrates, electrodes, and sensing composites. Successfully designed microstructure not only can be used to increase the sensitivity of piezoresistive but also that of capacitive sensors when microstructured dielectric is applied. Different modules have been developed, including micromachined wafers, silk fabrics, and even plant leaves. During the fabrication process, sensing materials are simply poured onto the module and peeled off after partial or complete drying. The adhesion between processed
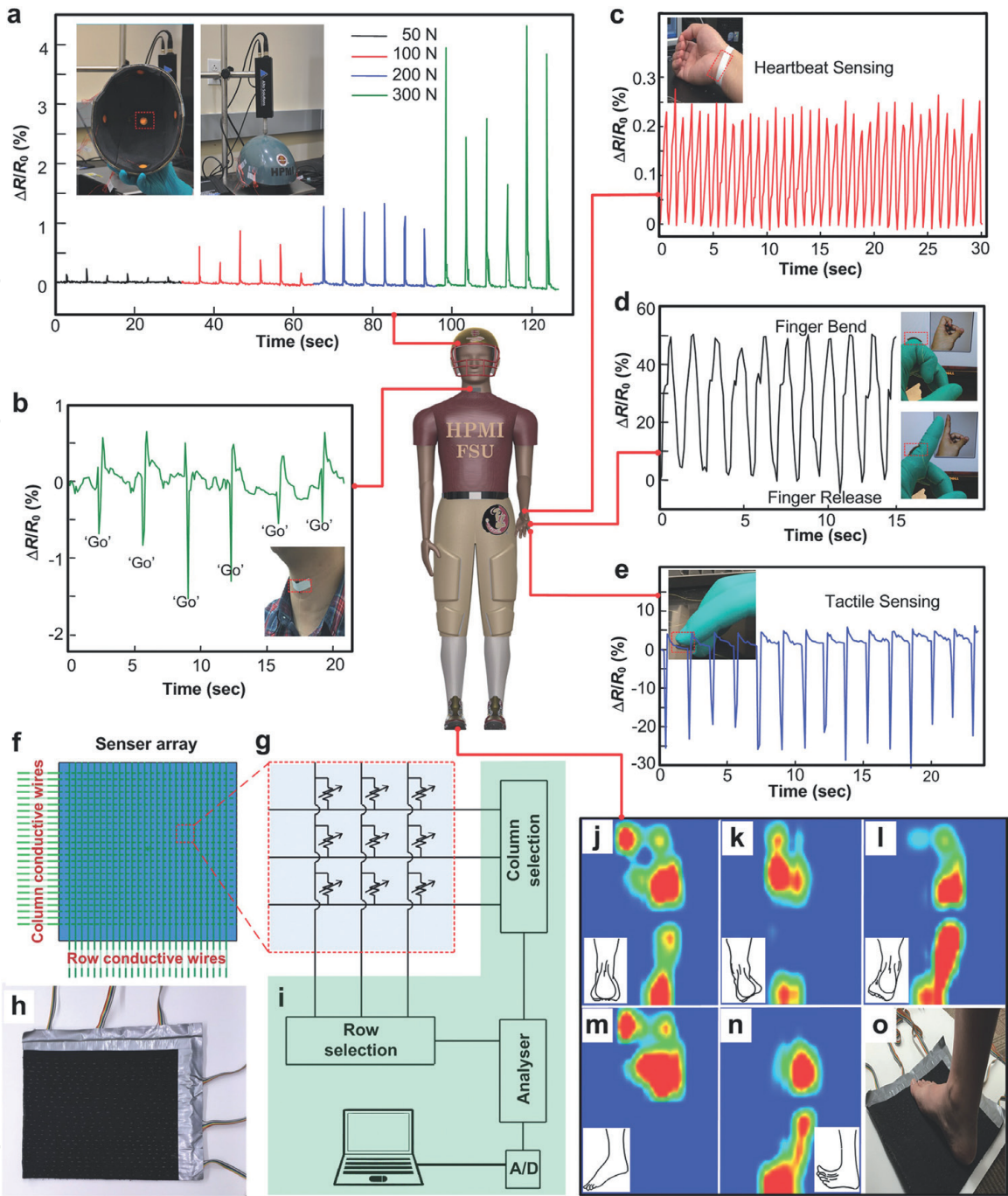
Figure 5.
Health motion monitoring with CNT-based strain sensor: (a) impact pressure, (b) muscle movement, (c) heartbeat, (d) finger motion, (e) finger touch, (f) schematic diagram of sensor array, (g) magnified view of the sensor array, (h) optical photograph of a fabricated sensor array containing 25 × 25 pixels, (i) circuit schematic of the sensor matrix (j–n) foot pressure pattern, and (o) strain sensor attached on the human right foot.

material and module is the most important parameters for this technique, which can be adjusted by necessary pretreatment and sophisticated geometric design.

3D printing is the best candidate for developing 3D constructions and has gained great popularity due to its powerful ability [66]. If the sensing materials are well prepared, arbitrary structures can be printed with 3D printing with adjustable resolution, even lower than 0.1 μm. For instance, A three-layer sensor has been fabricated in a single step by 3D printing, which originally requires multiple steps by using traditional method, including micromolding, laminating, and infilling. Wearable pressure sensor has also been realized by a multimaterial, multiscale, and multifunctional 3D printing approach. The size of this sensor is 3 × 3 mm in area and 1.2 mm in height [67].

5. Applications of wearable electromechanical sensor

Wearable electromechanical sensors can basically detect mechanical signals including pressure and strain. Applications that require monitoring pressure and strain are theoretically can realized by it. Until now, monitoring of human motion and health, speech recognition, gesture recognition, human machine interaction, acoustic waves detection, and even disease diagnosis have been demonstrated, which would be discussed in the following.

5.1 Human motion monitoring

When wearable electromechanical sensor is mounted on the skin or integrated with textiles, it can real-time monitor human motions including hand, limb, foot, face, and throat. Subtle deformations induced by body activities such as blood pulse flow and respiration, and large deformations related to the body movements such as finger and knee bending can be readily detected. **Figure 5** shows the human motions in daily life detected by CNT-coated auxetic foam strain sensor (AFS) [68]. As **Figure 5a** shows, the foam sensors performed well by dependably detecting the timing, frequency, and magnitude of the impact event and outputting signals in sharp spikes corresponding to the impact events. **Figure 5b** shows the monitoring of the muscle movement during speech by attaching a foam sensor onto a person's neck. When the person repeatedly says the simple words "go," stable signals can observe which timing and pattern corresponded well with the vocal events. Moreover, the wrist pulse has also been successfully monitored by the AFS (**Figure 5c**). A typical pulse waveform is obtained, and the pulse frequency of 76 beats min^{-1} can be calculated. It can also be used to transfer the human intentions of pressing buttons and switches by attaching the AFS directly to the fingertip (**Figure 5d**). **Figure 5e** demonstrates that the AFS can control gesture by wearing on the finger joint because the signal of the foam sensor one by one corresponds to the gesture. **Figure 5f** and **g** shows the schematic and a photograph of the sensor matrix, respectively. **Figure 5h** illustrates the sensing system and a simplified electrical schematic that scan the intersecting points of the sensor's rows and columns and measure the resistance at each crossing point. The plantar pressure distribution can be successfully analyzed with AFS matrix, further extending its fields of application ranging from sports performance and injury prevention to prosthetics and orthotics design. For further example, **Figure 5j–n** shows the various barefoot pressure distributions applied by a human right foot (**Figure 5o**), including neutral position, pronation, supination, plantar flexion, and dorsiflexion, which is displayed by the colored contour maps. The in-shoe plantar pressure measurement can also be finished by simply inserting AFS matrix into shoes. It can be anticipated that wearable electromechanical sensor can find a wide range of applications in human motion monitoring, body pressure distribution, and even adjusting sitting posture.

5.2 Human health monitoring

Human health monitoring is based on the continuous monitoring of human motions, especially the pulse and respiration. Wearable electromechanical sensor attached on wrist and chest can be used to detect the pulse and respiratory rate. **Figure 6** shows that graphene film strain sensor can exactly monitor people's pulse and breath rate. Strain sensor are attached on a person's wrist or chest for real-time recording of pulse and respiratory rate signals (**Figure 7a**) [69]. **Figure 6b** shows the collected pulse and respiratory signals, where each cycle represents a pulse or
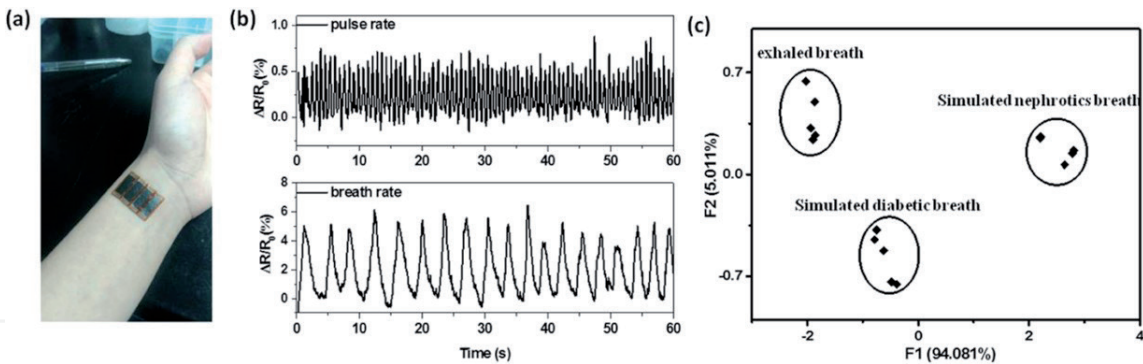
Figure 6.
Health monitoring with graphene strain sensor. (a) Photograph of strain sensor mounted on the human wrist, (b) normalized resistance changes of the strain sensor when monitoring wrist pulses and respiratory rate, and (c) PCA analysis of exhaled breath of simulated nephrotic patients, diabetic patients, and healthy people.
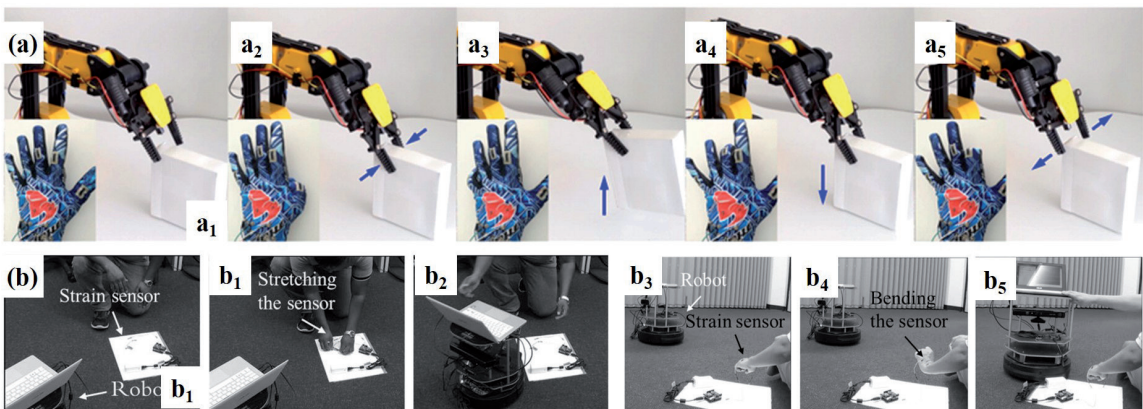

Figure 7.
Piezoresistive sensors for human-machine interfaces: (a) smart gloves and (b) robotic controlling.

breath. The valleys correspond to the shrinking of the chest, and peaks represent the stretching of the chest. Then, the pulse and breath rates can be estimated to be about 76 and 19 in 60 s, respectively. Three kinds of exhaled breath (simulated diabetic breath, simulated nephrotic breath, and the breath of healthy individuals) are investigated. The obtained response data are analyzed, and the results are displayed in **Figure 6c**. It can be observed that the three breath samples are clearly different. The exhaled breath samples are categorized into three distinguishable clusters without any overlap, which correspond to healthy individuals, simulated diabetic patients, and simulated nephrotic patients, respectively. This demonstrates that wearable strain sensor has high potential for human health monitoring and even the diagnosis diseases.

5.3 Speech recognition

Speech recognition is also based on the monitoring of human motions. When the wearable electromechanical is attached on the throat, it could record muscular movements in order to collect and recognize speech sounds. This is permitted by the fact that the throat muscle exhibits different degrees of stretching or shrinking strains when speaking different words. Due to the tiny changes caused by throat motion, the strain sensor used in speech recognition should have high sensitivity. The GF of GWF strain sensor can be as high as 10^3 with 2–6% strains, 106 with higher strains (>7%), and ~35 with a minimal strain of 0.2%, which is suitable for this application. The results show test signal waveforms of all 26 english letters [70]. As expected, the waveforms are unique and repeatable for all letters. Since each

individual speech organ is different, people can easily distinguish whether a given voice comes from the same person. This demonstrates that wearable electrome- chanical can be used in speech recognition.

5.4 Human-machine interface

Human-machine interfaces and robotic remote controlling are greatly beneficial in surgery or a highly risky work that requires the replacement of robotics. The electromechanical sensor used in human-machine interface is typically mounted on body joint, which are normally bended or stretched at large degree of deformations; thus, high stretchability (>50%) is required. The robotic controlling is demonstrated in **Figure 7**, and the wearable strain sensors are based on the hybrid of polyaniline and gold nanowires for a smart glove [9]. The sensor-based-smart glove is used to control the movement of a robot through wireless signals (**Figure 7a**). The robot is at relaxed state (a_1) and works as an arm that can clamp (a_2), lift up (a_3), put down (a_4), and release (a_5) an object based on different postures of human fingers as wearing the sensor. **Figure 7b** reveals the remote control on the robot movement by a strain sensor based on graphene. As can be seen in this figure, b_1 and b_4 demon- strate the robot at the relaxed state. As the strain sensor is stretched or bended, the robot starts working (b_2 and b_5) and moves to the controller (b_3 and b_6).

6. Conclusion and outlook

In this chapter, we discuss the working mechanism, fabrication methods, and applications of wearable electromechanical sensors. Piezoresistive sensor attracts more attentions due to its clear structure, mechanism, fabrication methods, and low cost. High sensitivity and stretchability have been achieved simultaneously. However, the stability and linearity are still limited for resistive-type sensor. Moreover, mass production with low cost is still a challenge. One strategy to reduce cost is developing novel fabrication methods, which can readily build high- performance sensor. Many applications have been demonstrated in a qualitative way by using strain or pressure sensor. However, the practical application needs more quantitative analysis, which requires further investigations.

Acknowledgements

Conflict of interest

There is no conflict of interest between authors.

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