Organic Transistor-Based Chemical Sensors for Wearable Bioelectronics

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CONSPECTUS: Bioelectronics for healthcare that monitor the health information on users in real time have stepped into the limelight as crucial electronic devices for the future due to the increased demand for "point-of-care" testing, which is defined as medical diagnostic testing at the time and place of patient care. In contrast to traditional diagnostic testing, which is generally conducted at medical institutions with diagnostic instruments and requires a long time for specimen analysis, point-of-care testing can be accomplished personally at the bedside, and health information on users can be monitored in real time. Advances in materials science and device technology have enabled next-generation electronics, including flexible, stretchable, and biocompatible electronic devices, bringing the commercialization of personalized healthcare devices increasingly within reach, e.g., wearable bioelectronics attached to the body that monitor the health information on users in real time. Additionally, the monitoring of harmful factors in the environment surrounding the user, such as air pollutants, chemicals, and ultraviolet light, is also important for health maintenance because such factors can have short- and long-term detrimental effects on the human body. The precise detection of chemical species from both the human body and the surrounding environment is crucial for personal health care because of the abundant information that such factors can provide when determining a person’s health condition. In this respect, sensor applications based on an organic-transistor platform have various advantages, including signal amplification, molecular design capability, low cost, and mechanical robustness (e.g., flexibility and stretchability).

This Account covers recent progress in organic transistor-based chemical sensors that detect various chemical species in the human body or the surrounding environment, which will be the core elements of wearable electronic devices. There has been considerable effort to develop high-performance chemical sensors based on organic-transistor platforms through material design and device engineering. Various experimental approaches have been adopted to develop chemical sensors with high sensitivity, selectivity, and stability, including the synthesis of new materials, structural engineering, surface functionalization, and device engineering. In this Account, we first provide a brief introduction to the operating principles of transistor-based chemical sensors. Then we summarize the progress in the fabrication of transistor-based chemical sensors that detect chemical species from the human body (e.g., molecules in sweat, saliva, urine, tears, etc.). We then highlight examples of chemical sensors for detecting harmful chemicals in the environment surrounding the user (e.g., nitrogen oxides, sulfur dioxide, volatile organic compounds, liquid-phase organic solvents, and heavy metal ions). Finally, we conclude this Account with a perspective on the wearable bioelectronics, especially focusing on organic electronic materials and devices.

1. INTRODUCTION

Continuous and selective detection of chemicals in body fluids or gaseous pollutants, without invasive and expensive analytical
tools, is an appealing solution for future healthcare applications. Inorganic, silicon-based electronic devices have been extensively developed for applications in the bioelectronics. However, because electrons are mostly used as charge carriers in electronic devices, they often exhibit limitations when applied to biological systems, which mostly rely on ionic transport instead of electrons. In recent decades, to combine biology with conventional electronics and convert biological signals to electronic signals and vice versa, conducting and semiconducting organic materials have emerged as excellent tools enabling both electronic and ionic conduction. Also, they can be easily synthesized to include specific receptors or anchoring sites for highly sensitive and selective chemical detection. Thanks to these advantages, organic electronics has become a general platform for biological recording and applications in life sciences and healthcare system.

In 1984, the first use of organic materials as an active component of transistors was reported by White et al., who investigated the conductivity changes in polypyrrole by electrochemical doping. After that, Koezuka et al. developed the first solid-state organic field-effect transistor (OFET) by using polythiophene as an active material. Since then, organic transistors have emerged as a feasible solution for high-performance chemical sensors by converting chemical information into electrical signal. Organic transistors can operate at low voltages, satisfying operation in aqueous environments and applications in inexpensive chemical or biological sensors. Transistor-based sensors have the combined device structure of a sensor and an amplifier, in which imperceptible electrical changes induced by specific analytes can lead to a pronounced change in channel current, resulting in high sensitivity compared with other types of sensors. Also, transistor-based sensors show various advantages, including fast response, feasibility for miniaturization, easy processing, and high throughput. In particular, organic transistor-based sensors are considered a key platform for flexible and wearable sensors for “point-of-care” health monitoring by minimizing the discomfort of wearing.

In this Account, we review the status of the development of organic transistor-based detection of not only chemicals excreted from the body in the form of biofluids and exhaled breath, but also those detrimental to health. Reflecting the increasing demand for wearable bioelectronic chemical sensors that directly interface with living and moving human bodies, we specifically discuss the application of organic transistor-based chemical sensors to wearable electronics (Figure 1).

2. OPERATION PRINCIPLES OF TRANSISTOR-BASED CHEMICAL SENSORS

Organic transistors have been spotlighted as a tool of choice for wearable bioelectronics applications involving chemical sensors, particularly due to their biocompatibility and easy integration into portable and wearable electronic devices. The main advantage of organic transistor-based sensors is higher sensitivity than two-terminal-based sensors, due to the signal amplification and control ability by modulating the gate voltage. In this section, the working principles of chemical sensors based on two types of organic transistors, conventional OFETs in thin-film structure and organic electrochemical transistors (OECTs), are introduced.

A typical OFET is composed of three electrodes, a gate dielectric layer, and an organic semiconductor film. The channel current flows between the source and drain electrodes, where charge carrier transport occurs near the interface between the semiconductor and dielectric layers. This can be controlled by modulating the voltage applied to the gate electrode \(V_G\), resulting in field-effect behaviors. The channel current \(I_{DS}\) of OFETs can be derived by the following equations:

\[ I_{DS} = \frac{W}{L} \mu \frac{C}{C} (V_G - V_T) V_{DS}, \quad V_{DS} < V_G - V_T \]

\[ I_{DS} = \frac{W}{2L} \mu \frac{C}{C} (V_G - V_T)^2, \quad V_{DS} > V_G - V_T \]

where \(W\) and \(L\) are the width and length of a channel, \(\mu\) is the field-effect mobility, \(C\) is the capacitance of the gate insulator, \(V_T\) is the threshold voltage of the device, and \(V_{DS}\) is the applied voltage between drain and source electrodes. When OFETs are used as chemical sensors, analyte diffusion through semiconductor grain boundaries can change \(\mu\), and the doping effect can change \(V_T\), affecting \(I_{DS}\). In most OFET-based sensors, the semiconducting layers are exposed to the target analytes. In this case, the organic semiconductor layer acts as both the electronic transport material and the chemical sensing layer (Figure 2a).

In the case of OECTs, an electrolyte medium is used instead of a dielectric layer, and the gate electrode is connected to the

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**Figure 1.** Wearable bioelectronics applications of organic transistor-based sensors for noninvasive detection of chemicals excreted from the body and exogenous chemicals detrimental to health.
active layer via an electrolyte, in which ionic species penetrate the active layer and modulate $I_{\text{DS}}$ via changes in the doping state (Figure 2b).\textsuperscript{14,15} The changes in electrical current of an OECT can be derived by the following equations:\textsuperscript{8}

$$I_{\text{DS}} = \frac{qnp_0 t W}{LV_p} \left( V_p - V_G^{\text{eff}} + \frac{V_{\text{DS}}}{2} \right) V_{\text{DS}},$$

$$|V_{\text{DS}}| \ll |V_p - V_G^{\text{eff}}|,$$

$$V_p = \frac{qnp_0 t}{C_i}, \quad V_G^{\text{eff}} = V_G + V_{\text{offset}}$$

where $q$ is the electric charge, $p_0$ is the initial carrier density when $V_G$ is not applied, $t$ is the thickness of the active layer, $V_p$ is the pinch-off voltage, $V_G^{\text{eff}}$ is the effective gate voltage, and $V_{\text{offset}}$ is the offset voltage related to the potential drop of the OECT. Most OECT-based chemical sensors operate via changes in potential drop across the gate/electrolyte or electrolyte/channel interfaces, with exposure to the analytes. Both OFETs and OECTs utilize a method of introducing functional receptors into the device structure for selective and sensitive chemical detection.\textsuperscript{16} Three approaches—active/receptor bilayer structure, assembly of active layer-receptor, and active/receptor composite—are commonly used for receptor implantation (Figure 2c).

3. RECENT PROGRESS IN TRANSISTOR-TYPE ORGANIC CHEMICAL SENSORS FOR HEALTH SELF-MONITORING

3.1. Sensors Detecting Chemicals from the Human Body

The history of biosensors for detecting chemicals from the human body began in 1962 with Clark et al., who successfully monitored glucose concentrations in blood by employing glucose oxidase (GOx).\textsuperscript{17} Since then, a variety of chemical sensors that monitor specific chemicals secreted from the human body have been developed for biological and medical diagnostic applications.\textsuperscript{18} Here, we focus on wearable healthcare systems satisfying the demand for easy, rapid, inexpensive, and portable chemical detection (Figure 3). Subsection 3.1.1 focuses on wearable organic transistor-based sensors for detecting chemicals in bodily excreted products, and in subsection 3.1.2 other kinds of wearable sensors are described.

3.1.1. Chemicals in Sweat, Saliva, Urine, and Tears

Sweat, a representative noninvasive biofluid, is a useful analyte because it can be detected in real time without human intention through sensors worn on the body. Because blood glucose is diffused at a certain rate into sweat, the sweat glucose sensor is a promising noninvasive diagnostic platform for diabetes.

You and Pak demonstrated a graphene FET enzymatic wearable sensor that can detect glucose.\textsuperscript{18} Using silk fibroin film as both an enzyme immobilization matrix and substrate, the sensor could be attached to human skin, such as the wrist (Figure 4a). The GOx-immobilized sensor was selectively responsive to glucose in sweat at a range of 0.1–10 mM due to the enzymatic reaction.

Minami et al. reported an enzyme-based OFET sensor with an extended gate electrode on a flexible polyethylene
naphthalate (PEN) substrate for detecting lactate, which is a useful biomarker of anaerobic metabolism in circulatory failure. To directly expose the sensor to sweat, the extended gate, on which lactate oxidase (LOx) was immobilized, was used as a sensing unit (Figure 4b). The drain current changed upon stepwise changes in lactate concentration, due to the enzymatic redox chain reaction on the extended gate electrode. The detection limit was estimated to be as low as 66 nM.

Promisingly, highly sensitive biological sensors can be used to sense various chemicals in saliva for noninvasive healthcare. Liao et al. developed OECT-based uric acid (UA) sensors by modifying platinum (Pt) gate electrodes with positively charged polyaniline (PANI) and negatively charged Na grapheme flake bilayer films with the enzyme uricase (UOx) (Figure 4c). The fabricated sensors demonstrated outstanding selectivity and a detection limit of \( \sim 10 \text{ nM} \) UA in saliva by blocking the interfering materials using a charged bilayer film. Also, long-term stable performance was observed during 1000 times of bending.

Selective sensing of salivary glucose was reported with an OECT-based sensor. Ji et al. modified a gate electrode with GOx and poly(n-vinyl-2-pyrrolidone)-capped Pt nanoparticles (NPs) and used a microfluidic channel. High sensitivity down to \( 10^{-7} \text{ M} \) as well as enhanced glucose selectivity was achieved by depositing a negatively charged Nafion layer on the gate electrode, which suppressed the diffusion of other compounds with electrostatic interactions. A prototype portable sensor for real-time glucose sensing was demonstrated (Figure 4d).

Conventional analytical methods for detecting drugs require long operation time, expensive equipment, and complex experimental procedures. Jang et al. successfully demonstrated an easy, fast, and inexpensive detection method for amphetamine-type stimulant (ATS) via the synergistic effects of selective supramolecular chemistry and OFET platform. Cucurbit[7]uril (CB[7]) derivatives with the cavity and carbonyl-laced portals that can interact with ammonium groups on ATS by ion–dipole interactions were deposited on a 5,5′-bis(7-dodecyl-9H-fluorene-2-yl)-2,2′-bithiophene (DDFTTF) layer (Figure 5a). The amperometric OFET sensors showed a detection limit for ATS as low as 1 pM in DI water and 1 nM in real urine. Furthermore, a wearable wristband-type drug sensor was demonstrated for stable and repeatable real-time detection of ATS in urine.
Yang et al. developed OECT-based biosensors in which Cr/Au and poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate) (PEDOT:PSS) layers were deposited on nylon fibers (Figure 5b). The fabricated Cr/Au/PEDOT:PSS-coated fibers were functionalized and woven together, resulting in flexible and stretchable fabric biosensors. Upon exposure to biological interfaces, such as phosphate-buffered saline (PBS) solution and artificial urine, the woven OECT sensors showed superior selectivity and low detection limits for glucose (30 nM), dopamine (10 nM), and UA (30 nM).

Kim et al. reported a graphene FET-based wearable soft contact lens for wireless detection of glucose in tears, with high sensitivity. The graphene–AgNWs hybrid served as both the antenna coil and source/drain electrodes of an graphene-based FET. To provide reliable and comfortable vision, all the components were sufficiently transparent and a real soft contact lens was used as the template (Figure 5c). GOx was immobilized on the graphene channel, allowing selective binding with glucose. Integrated into a resistance, inductance, and capacitance (RLC) circuit, the sensor exhibited real-time in vivo glucose detection when attached directly to a living rabbit eye.

3.1.2. Chemicals in Breath and Other Internal Body Fluids. Analysis of breath chemicals is a state-of-the-art method in healthcare and disease diagnostics. Various methods have been introduced to detect chemicals in the breath. For example, ammonia in the breath can be used as a biomarker of renal failure. Zhang et al. developed a nanoporous-structured OECT-based ammonia sensor. Nanopores were introduced into poly(diketopyrrolopyrrole-thiophene-thieno[3,2,b]thiophene-thiophene) (DPP2T-TT) thin film, providing direct access to thereactive sites (Figure 5d). The sensor showed a detection limit of ammonia below 1 ppb and a reduced response time at the hundred-millisecond time scale. Moreover, the sensor fabricated on a flexible substrate showed real-time breath-sensing performance in the range from 10 ppb to 10 ppm.

Lactate, typically produced by increased metabolic activity, has been suggested as a biomarker for cancer. Braendlein et al. reported an in vitro OECT circuit for sensitive and accurate metabolite detection in highly complex cell culture media. They adopted the Wheatstone bridge sensor platform for inherent background subtraction. Chitosan–ferrocene complex was utilized with nonspecific protein on the reference OECT and LOx was attached to the sensing OECT (Figure 5e). They estimated the detection limit of lactate to be ~10 μM. In addition, chrono-potentiometric recording was conducted with exposure to complex cell media of malignant samples. This was the first demonstration of a miniaturized sensor circuit for cancer diagnosis.

Thus, far, even though many biosensors have used enzyme immobilization methods, these approaches have been found to be expensive, and have slow responses and complicated fabrication processes. Solving these problems, Jang et al. demonstrated OFET-based biosensors that can detect the acetylcholine (ACh+), using biological recognition elements, a cucurbit[6]uril (CB[6]) derivative. They focused on combining the high sensitivity of the transistor configuration and the high selectivity of synthetic receptors. The CB[6]-derivative film was deposited on the surface of a water-stable semiconductor (DDFTTF) layer. The fabricated biosensor showed a low ACh+ detection limit of ~1 pM with excellent discrimination over choline (Ch+) in real-time. Furthermore, flexible sensor was fabricated using indium tin oxide (ITO)-coated PEN substrates (Figure 5f), which could detect ACh+ with an identical detection limit (~1 pM) in PBS solution under low-voltage operation.
3.2. Sensors Detecting Chemicals Detrimental to the Human Body

In the applications of chemical sensors in bioelectronics, the detection of chemical compounds that significantly affect the human body is a crucial application area for wearable electronics. With rapid industrialization, humans have been ubiquitously exposed to various harmful pollutants.36 These pollutants seriously affect the health of human respiratory, cardiovascular, nervous, and other systems. Therefore, the demand for real-time monitoring devices that detect pollutants in the atmosphere and the water has risen. In this section, we introduce the recent progress in organic transistor-based chemical sensors that detect chemicals detrimental to human health, including volatile organic compounds (VOCs); gaseous pollutants such as nitrogen oxides (NOx) and sulfur dioxides (SO2); liquid-phase organic solvents; and heavy metal ions. Specifically, subsection 3.2.1. deals with chemical sensors for VOCs and subsection 3.2.2. focuses on sensors that detect gaseous pollutants. Then, recent progress in organic transistor-based chemical sensors that detect liquid-phase organic solvents and heavy metal ions is described in subsections 3.2.3. and 3.2.4., respectively.

3.2.1. Volatile Organic Compounds (VOCs). With the increasing use of chemicals in various products, the generation of harmful chemical vapors is receiving considerable attention as a rising source of chemical pollutants. VOCs are classified as organic compounds that have a boiling point between 50 and 260 °C, including various chemicals such as benzene, ammonia, and acetone.35 Because these chemicals have been used in many commercial products for various purposes, VOCs can be generated from any products containing them, even from most household products such as paints, detergents, and furniture.35 Due to the growing threat posed by VOCs which can affect health (e.g., irritation, headaches, nausea, liver/kidney damage, and central nervous system damage), research on OFET-based sensors for the detection of VOCs is gradually increasing.

The development of a high-performance pentacene-based OFET-type methanol sensor was reported by Kang et al.33 Sequential evaporation of m-bis(triphenylsilyl) benzene (TSB3) as a rubbery small-molecule dielectric layer and pentacene as the active material on n-octadecyltrimethoxysilane (OTS)-treated SiO2 substrate produced a macroporous pentacene/TSB3 layer. For chemical sensing applications, macroporous pentacene/TSB3-based OFET sensors showed high sensitivity toward methanol vapor, with facilitated diffusion of analyte molecules into the channel region via the macropores (Figure 6a). Cheon et al. reported graphene oxide (GO)-functionalized OFET-based sensors for the detection of VOC gas.34 DPP-selenophene-vinylene-selenophene (DPP-SVS) doped with functional GO was utilized as an active GO-OFT structure for the detection of various analytes including ethanol, acetone, and acetonitrile. Due to its many polar functional groups and high surface-to-volume ratio, GO functioned as a gas-adsorbing dopant. Compared with OFET sensors based on pristine DPP-SVS, GO-functionalized OFET-based sensors exhibited highly enhanced gas sensing properties by showing significant changes in on-current and threshold voltage.

Nkeita-Yawson et al. reported the development of highly sensitive and flexible ammonia sensors based on organic printed transistors.35 Flexible OFET sensors based on an ultrathin film of a poly(4-(4,4-bis(2-ethylhexyl)-4H-silolo[3,2-b:4,5-b′]dithiophen-2-yl)-7-(4,4-bis(2-ethylhexyl)-6-(thiophen-2-yl)-4H-silolo[3,2-b:4,5-b′]dithiophen-2-yl)-5,6-difluorobenzol[c] [1,2,5]thiadiazole) (PDFDT) active layer were fabricated and utilized for ammonia vapor sensing. High sensitivity was demonstrated with hydrogen bonding and electrostatic interactions between fluorine on PDFDT and ammonia. Fabricated sensors showed significant changes in electrical properties under the condition of a small amount of ammonia gas exposure as low as 1 ppm (Figure 6b). A study on the sensing mechanism indicated that hydrogen bonding and electrostatic interactions between the fluorinated aromatic backbone of PDFDT and ammonia molecules lowered the highest occupied molecular orbital (HOMO) level of PDFDT, which produced more hole traps to hinder charge transport in the channel.

Lee et al. demonstrated the development of an OFET-type flexible chemical sensor array based on semiconducting polymers with high chemical robustness.36 DPP-selenophene copolymers with siloxane-terminated chains (PTDPPSe-SiC4)-based OFETs showed high insolvibility in chemical solvents and could operate after immersion in various solvents for 1 day. Based on its chemical robustness, semiconducting PTDPPSe-SiC4 was applied to a conventional photolithography process for the fabrication of a flexible OFETs patterned array. The fabricated OFET’s array operated well after undergoing the harsh photolithography process and showed long-term stable acetone vapor sensing behaviors (Figure 6c).

3.2.2. Gaseous Pollutants. Air pollution negatively impacting human health is primarily due to the combustion of fossil fuels used for energy generation and transportation. Among the various gaseous products from fuel combustion, NOx and SO2 are the main causes of human health problems such as respiratory problems, premature death, and preterm birth.37 For organic transistor-based NOx sensors, various approaches to enhance the sensitivity of the sensors have been proposed. Huang et al. increased the sensitivity of the NO2 sensors by modifying the interface between the semiconducting layer and dielectric layer using UV-ozone (UVO) treatment.38 The UVO-treated sensors showed ~400 and ~50 times higher sensitivity toward 30 and 1 ppm of NO2, respectively (Figure 6d). Zang et al. suggested OFET-based gas sensors for detecting gas analytes, including NOx based on chemical reactions between gas analytes and semiconductors (Figure 6e).39 Interestingly, in comparison with the distinct signal change of a sensor toward 10 ppm of NO2, no obvious signal responses were observed toward 10 ppm of SO2 gas, likely due to the rather weaker electron withdrawing ability of SO2 than of NO2, suggesting the selectivity of the sensors. For SO2 sensors, Liu and co-workers demonstrated gas dielectric transistors based on single crystalline metal phthalocyanines (i.e., CuPc nanowire and ZnPc nanobelts), which have been widely utilized as the sensing layer in gas sensors due to their high thermal and chemical stability.40,41 OFET-based SO2 sensors with gas dielectrics showed higher sensitivity than those with a solid-state dielectric (i.e., PMMA).30 Gas dielectric SO2 sensors with CuPc nanowire exhibited a sensitivity and detection limit of 119% and 0.5 ppm, respectively (Figure 6f). For ZnPc-based SO2 sensors, a detection limit of 50 ppb and signal response of 910% with 10 ppm of SO2 gas were observed.41

3.2.3. Liquid-Phase Organic Solvents. Although gaseous pollutants in the atmosphere mainly affect human health through respiration or skin contact, liquid-phase chemicals are also noxious to the human body through ingestion or skin absorption. As public concerns about water contamination have increased, the demand for portable or wearable sensors that detect liquid-phase organic solvents has risen rapidly. There is a substantial limitation of...
OFET-based chemical sensors have poor resistance toward liquid-phase organic solvents, hindering further applications in detecting liquid-phase solvents. Thus, there have been various approaches to enhance the solvent resistance of organic semiconductors, such as passivation and cross-linking of semiconductors.36,42,43

The chemical cross-linking of the semiconducting layers of the sensors could enhance solvent resistance toward liquid-phase organic solvents. Lee et al. reported solvent-resistant OFET-based chemical sensors that detect liquid-phase organic solvents via direct injection.42 They utilized specially functionalized polymer semiconductor, P3HT-azide, where azide helps an efficient photo-cross-linking to generate OFETs with high solvent resistance. Furthermore, the surface of the P3HT-azide layer was modified with the container molecule, calix[8]arene (C[8]A), in order to enhance the sensing performance of the sensors (Figure 7a). The surface-functionalized P3HT-azide sensors showed distinct sensing responses toward various liquid-phase organic solvents (Figure 7b). To demonstrate the potential of the sensors as portable and wearable sensors, a flexible sensor was fabricated using a flexible substrate (PEN) and a polymer dielectric (cross-linked poly-4-vinylphenol [PVP]), showing an identical sensing trend toward liquid-phase organic solvents (Figure 7c).

3.2.4. Heavy Metal Ions. Environmental concerns about heavy metal ions in water waste and the ocean have risen with increasing industrialization. High levels of exposure to heavy metals generally increase serious health risks, such as cancer and nervous system damage.46 Knopfmacher et al. developed water-stable OFET-based chemical sensors to detect mercury ions (Hg\textsuperscript{2+}) in the marine environment (Figure 7d).44 The surface of the semiconducting layer (P3HT-Si) was modified with the DNA-functionalized Au NPs to introduce selectivity of the sensors toward Hg\textsuperscript{2+}, showing a distinct signal response under Hg\textsuperscript{2+} exposure compared with other heavy metal ions (i.e., Zn\textsuperscript{2+} and Pb\textsuperscript{2+}, Figure 7e). Flexible Hg\textsuperscript{2+} sensors were also demonstrated using polyimide and cross-linked PVP as the substrate and dielectric layer, respectively. Sudibya et al. demonstrated chemical sensors that detect various metal ions including Hg\textsuperscript{2+} and Cd\textsuperscript{2+} based on micropatterned reduced GO (rGO).45 After patterning the GO on the substrate, the GO was reduced to rGO and utilized as the sensing layer of the sensors (Figure 7f). The detection limits of the sensors toward Hg\textsuperscript{2+} and Cd\textsuperscript{2+} were both 1 nM.
In this Account, we introduced the working principles and recent progress in organic transistor-based chemical sensors, mainly focusing on the wearable organic sensors for detection of the chemicals from human body as well as harmful exogenous chemicals. Unlike most previously reported wearable sensors detecting physical movements, the new generation of wearable chemical sensors aims for real-time detection of health-related chemicals. The key features of wearable organic chemical sensors include enhanced flexibility, stretchability, biocompatibility, low-cost production, and point-of-care monitoring. With continuing development, organic bioelectronic devices are now complementing inorganic counterparts.

Although the electrical performance of organic electronics has caught up to or even exceeded that of amorphous silicon devices, there are still issues to be improved for commercialization, such as operational and environmental long-term stability. Various demonstrations of organic transistor-based chemical sensors have been reported for human health monitoring devices, as stated above. However, many issues still remain to be resolved for practical applications. For example, selectivity toward certain target analytes can easily be decreased in the real operating environment, because various surrounding factors may interfere with the sensing results. The long-term stability of the chemical sensors, inter alia, is a critical issue in organic sensors because the sensing layer is generally exposed to the external environment in order to interact with target analytes. In recent years, however, there has been considerable progress in enhancing the environmental robustness of organic electronics. As huge efforts have been made to understand the degradation mechanisms of organic electronic devices and improve their operational and environmental long-term stability, the potential applications of organic electronic devices in wearable electronic devices have increased accordingly. A number of strategies have also been developed to overcome such weaknesses by optimizing material design, device geometry, circuit design, and analytical method. Along with the development of the Internet of things (IoT) and big data technologies, wearable devices based on organic chemical sensors are envisioned to facilitate point-of-care health monitoring and enable saving lives and disease prevention with interdevice communication. Combined with the physical sensor technologies, wearable chemical sensors are expected to...
construct useful diagnostic data for human health monitoring and future nano—bio information technologies.

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