Chapter 9

Transistors as an Emerging Platform for Portable Amplified Biodetection in Preventive Personalized Point-of-Care Testing

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Additional information is available at the end of the chapter

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Abstract

The impressive improvement in biomolecular detection has gone from simple chemical methods to sophisticated high throughput laboratory machines capable of accurately measuring the complex biological components and interactions. In the following chapter, we focus our attention on transistor-based devices as an emerging platform for easy-to-use, portable amplified biodetection for preventive personalized medical applications and point-of-care testing. Electronic sensing devices comprise biosensors based on field-effect transistors (bio-FETs) and organic electrochemical transistors (OECTs). Transistor sensing devices can transduce electronic and ionic signals thereby creating an effective human-machine communication channel. In this chapter, we survey the progress done on the development of transistor innovative concepts to examine biological processes, i.e., biosensors integrated with textiles, flexible substrates, and biocompatible materials. Electrochemical and field-effect transistors can operate at low voltages possibly serving for highly sensitive, selective, and real-time sensing devices. The exploration of biosensors integrates different disciplines such as organic electronics, biology, electrochemistry, and materials science.

Keywords: bio-FETs, OECTs, transistor biosensors, electrolyte-gated transistors

1. Introduction

Biosensors are devices that incorporate biological sensing elements or bioreceptors to detect specific molecule/chemical analytes and produce a measurable signal (current, voltage, and

color change). Enzymes are protein molecules that act as biological catalysts that bind/react highly specific to certain molecules therefore enzymes can be excellent bioreceptors. The transducer is the mechanism by which selective binding of analytes and bioreceptors is converted into a measurable signal. Common transducers are based on optical, electrochemical, and electronic signal modulations upon addition of the analyte.

Electronic devices based on polymer electrodes, organic thin-film transistors, and organic light-emitting diodes can be interfaced with biological moieties, i.e., cells, microorganisms, proteins, oligonucleotides, small molecules, for medical applications, and environmental and food quality control [1–7]. The scope of bioelectronics is shown in **Figure 1** [7]. Transistors, polymer electrodes, and organic light-emitting diodes can be coupled with recognition element bioreceptors, i.e., enzymes, nucleic acids (DNA), or antibodies to selectively detect an analyte, i.e., proteins, complementary DNA chains, viruses, nutrients, hormones, collected from blood, urine, or saliva [8, 9]. Transistor-based biosensors can be divided into field-effect transistor biosensors (bio-FETs), discussed in Section 2 and organic electrochemical transistors (OECTs), discussed in Section 3.

1.1. Background of biosensors and main transducer mechanisms

Since the introduction of insulin as a treatment for diabetes, monitoring glucose in blood became an important issue. Glucose sensors are nowadays by far the most successful commercial application of biosensors with more than 85% share of a multibillion dollar market in the USA [8, 10]. For this reason, glucose biosensors lead the path toward the development of new biosensing concepts.

The main transducer mechanisms involve optical, electrochemical, and electronic processes.

1.1.1. Optical transducers

The early glucose biosensors were based on colorimetry. Colorimetry is a technique in which the color of a solution is taken as an index of the composition of the solution. The color of the

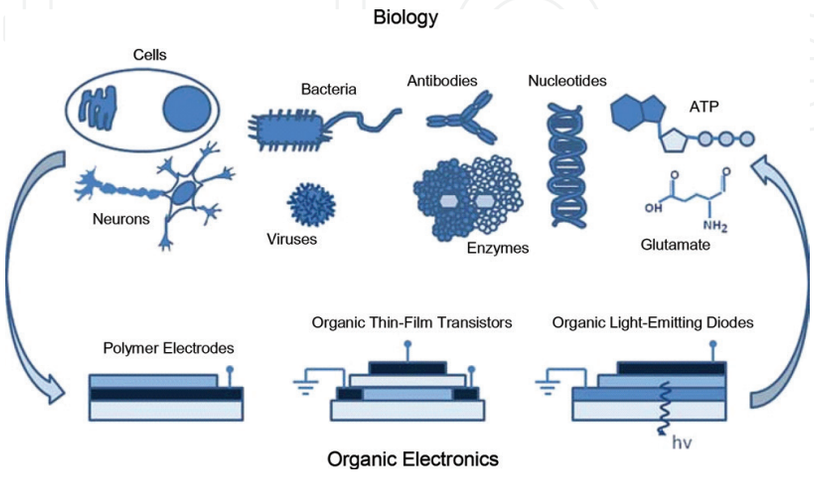

Figure 1. A cartoon showing the scope of organic bioelectronics [7]. Copyright © Materials Research Society 2010.

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solution can be modified by a chemical reaction and the concentration of its constituents can be revealed by comparison with a suitably prepared set of color standards in a color comparator [11]. The color comparison of the solution with its standard can be made visually or with a colorimeter. Colorimeters are devices that measure light absorption or reflection in the visible spectrum.

Most of the colorimetric methods to measure glucose in blood produce a color change by the reduction of an aromatic organic compound. The introduction of Dextrostix (Ames Co.) in 1964 provided a simple tool to measure blood glucose. Dextrostix is a cellulose strip with reagents that vary in color and intensity as a function of the amount of glucose present in the blood [12]. The reagents in Dextrostix are glucose oxidase, peroxidase, and a chromogen (i.e., tetramethylbenzidine). Glucose in blood samples is first oxidized into gluconolactone and eventually into gluconic acid [13]. This oxidation reaction is assisted by the enzyme glucose oxidase (GOx) biological catalyst. Oxygen dissolved in blood is necessary to carry this oxidation process and the by-product is hydrogen peroxide (H_2O_2). Hydrogen peroxide then reacts with the chromogen system to result in a specific color pattern specific to the glucose concentration in blood.

To avoid the operator subjective comparison of the strips by visual inspection, the color change in Dextrostix strips can be evaluated with a simple reflectance meter or with more sophisticated spectrophotometers.

Despite the simplicity of optical transducers, its early development was hampered by the requirement of bulky and costly optical equipment or by the intrinsic inaccuracy of visual inspection [14].

One possible solution may be replacing costly reflectance meters or spectrophotometers with software algorithms to enable digital cameras in smartphones quantitative screening when fast diagnosis is needed. Common CMOS cameras found in smartphones have successfully monitored glucose, protein, and pH in artificial urine with interphone repeatability and minimal operator intervention [15].

Silicon optical photonic-crystal waveguide arrays capable of sensing up to 128 compounds within a few millimeters of space with excellent signal-to-noise ratio have been recently reported [16, 17]. The operating principle is similar to that in surface plasmon resonance sensors that measure the change in reflectivity of a thin gold metal film deposited on a glass prism upon addition of the analyte. Photonic-crystals are optical structures in which a periodic modulation of the refractive index exists for a given material. Depending on the exact periodic modulation, a given bandwidth of light cannot be transmitted through such material thus defining a photonic band gap. A given bandwidth of light can be confined in the waveguide. Since the band gap is very sensitive to the refractive index at the crystal surface, when in contact with aqueous medium, biomolecules adhere to the crystal surface and modify the refractive index resulting in a shift in the band gap. This property can be utilized for biosensing.

Surface functionalization with a binding bioreceptor of the photonic-crystal can be employed to improve analyte selectivity. While the requirement of additional optical and electronic components makes photonic-crystal biosensors relatively complex they can be made ultracompact in a single integrated chip embedded in an easy-to-use and portable device.

1.1.2. Electrochemical transducers

Electrochemical sensors originally called "enzyme electrodes"emerged as highly sensitive, easy-to-use, portable, and user-friendly sensing devices. Electrochemical biosensors are nowadays the most widespread transducers for glucose sensing [18].

Electrochemical sensors measure electrochemical processes occurring in an electrode functionalized with enzyme bioreceptors immersed in an electrolyte solution containing the analytes. Electrochemical processes include the measurement of tiny changes of voltage (potentiometry), current (amperometry), or resistance/conductance (conductometry) specific to the presence of an analyte. Electrochemical sensors usually involve two electrodes, the working electrode and the counter electrode under which a voltage is applied. A third reference electrode can be employed to set and monitor the potentials vs an absolute reference value.

In the absence of the reference electrode, it is somewhat difficult to measure a small current change in a reproducible way, thus the reproducibility of electrochemical sensors is generally less accurate than those employing optical transducers [19]. However, because electrochemical sensors can be miniaturized and manufactured inexpensively, they actually dominate the biosensor market [10, 19, 20].

Electrochemical glucose sensors alike their analogue glucose optical sensors employ glucose oxidase (GOx) enzyme to bind the glucose molecule and facilitate its oxidation process. Oxygen and GOx oxidize glucose into gluconic acid and generate hydrogen peroxide as by-product. During this process, GOx is reduced and can be regenerated (oxidized) by adding ferricyanide which in turn is reduced into ferrocyanide. A metal electrode can regenerate ferrocyanide (reduced form) into ferricyanide. The reduction-oxidation cycle of glucose generates electrons at the metal electrode and can induce a spontaneous electric current (with no voltage applied) proportional to the glucose concentration. To obtain a quick glucose measurement, a change in the electrical current is measured as a function of a voltage applied between the electrolyte and the metal electrode.

Another popular redox system to detect glucose is based on enzyme glucose dehydrogenase as catalyzer and nicotinamide adenine dinucleotide, replacing oxygen to oxidize glucose [21, 22].

1.1.3. Electronic transducers

After the discovery of the enzyme electrode, ion-sensitive field-effect transistors (ISFETs), **Figure 2** (right), in which the gate electrode in a conventional metal-organic field-effect transistor (MOSFET), **Figure 2** (left), is replaced by an aqueous solution and a reference electrode, emerged to measure ionic species in electrochemical and biological environments. Double layers formed at the oxide electrolyte interface result in a different conductive state at the transistor channel proportional to the electrolyte ion concentration [23]. The gate oxide can be sensitive to specific ions similar to a glass electrode [24] or can be modified with a selective membrane or molecular receptors to filter specific ions [25, 26]. ISFETs are not strictly biosensors because they do not employ a biomolecular receptor as an active component to sense ions but they laid the foundation toward field-effect transistor biosensors (bio-FETs).

Electronic transducers include the field-effect transistor and the organic electrochemical transistor, reviewed in greater detail in the following sections.

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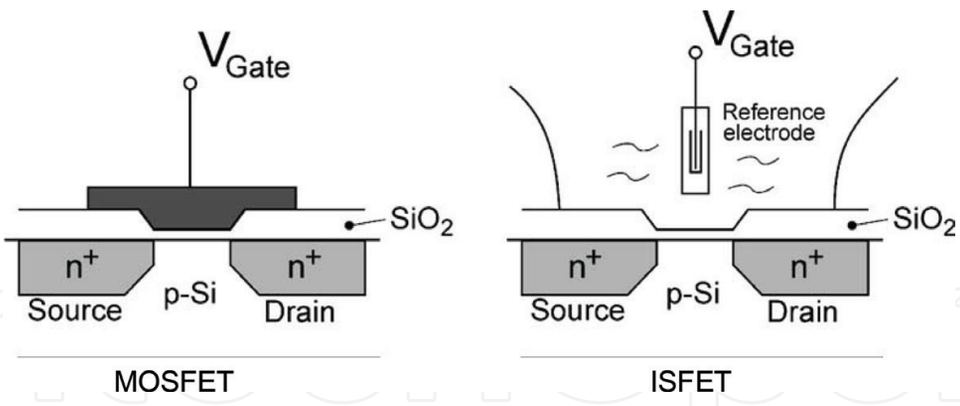

Figure 2. Schematic representation of MOSFET (left) and ISFET (right) typical structures [27]. Copyright © 2002 Elsevier Science B.V.

2. Field-effect transistor-based biosensors

ISFETs can be considered a close version of field-effect transistor-based biosensors (bio-FETs). Bio-FETs have the transistor semiconductor channel directly coupled with molecular bioreceptors sensitive to specific analyte molecules in the electrolyte without an insulating layer and can be directly gated through an electrolyte medium or through a back gate/gate-insulator as shown in **Figure 3**. The molecular bioreceptors act as a filter that allows only one type of analyte to selectively interact with the semiconductor channel. The analyte of interest can bind covalently to a specific molecular bioreceptor and change the semiconductor doping state. The conductive state of the functionalized semiconductor channel can be tracked by measuring the transistor drain-source current as a function of the electrolyte analyte concentration and the voltages applied.

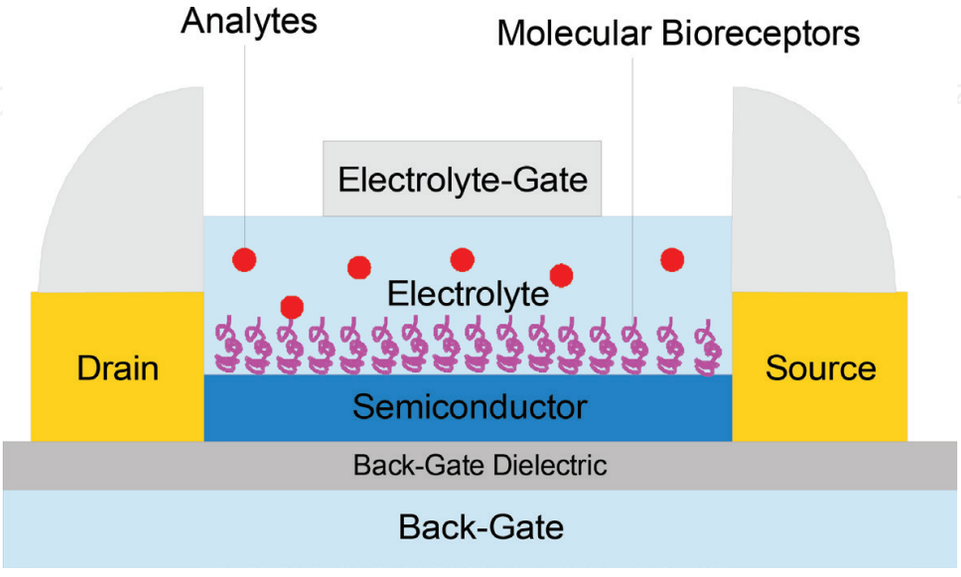

Figure 3. Schematic structure of field-effect transistor-based biosensors (bio-FETs).

Sensing mechanism may differ significantly when applied to different analyte molecules despite the common architectures among bio-FETs. As a thought experiment, a substance impermeable to ions that prevents the ionic doping of the semiconductor can be placed on top of the transistor channel. If the permeability of this substance changes following an interaction with a specific analyte allowing access of ions from the electrolyte into the semiconductor, the increased permeability of the barrier can be revealed by the modulation of the semiconductor conductive state as a function of a gate voltage [28]. Ions in proximity with the semiconductor can induce a doping/dedoping process by the field-effect or tune the effective energy barrier height required to inject charge carriers from the drain-source metal electrodes into the semiconductor. Because the chemical structure of the semiconductor remains unmodified, desorption of the analyte or ionic species from the semiconductor surface is possible for real-time reversible sensor detection.

Sensing DNA hybridization with FET devices is paving the way toward virus sensing and DNA disease prevention [29]. The working mechanism of DNA field-effect transistor sensors deserves attention. Carbon nanotube FETs typically operate as unconventional Schottky barrier transistors in which current modulation occurs primarily by tuning the contact resistance rather than the channel conductance [30].

Synthetic DNA hybridization consisting of random generated sequences and different oligo lengths (15 and 30 mer) was detected with bio-FETs made of gold drain-source electrodes functionalized with MercaptoHexanol and single-walled carbon nanotube channel material, **Figure 4** (left). Selective response to addition of complementary DNA was observed and almost no change occurred upon addition of phosphate-buffered saline (PBS) solution or mismatched DNA, **Figure 4** (right). MercaptoHexanol self-assembled monolayer provides a nice passivation on gold electrodes against nonspecific binding of mismatched DNA and provides ideal conditions for efficient hybridization with nearly 100% binding efficiency of analytes carrying complementary sequences. The formation of double stranded DNA on gold electrodes lowered the effective work function of gold facilitating charge carrier injection.

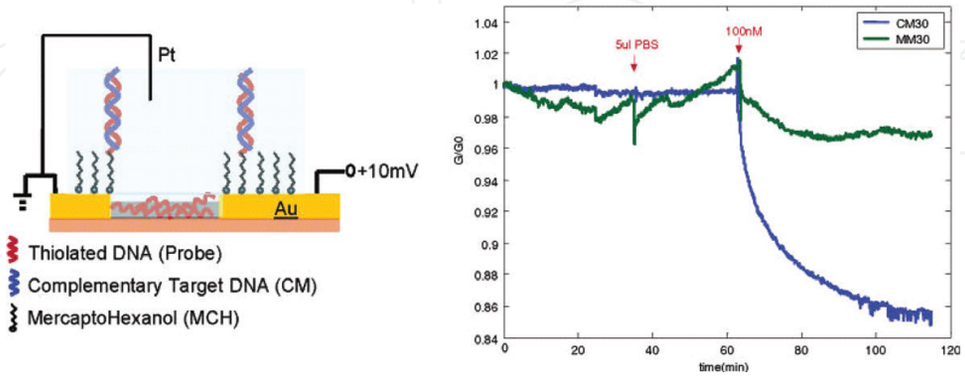

Figure 4. Schematic illustration of DNA FETs sensing device in operation (left) and real-time normalized conductance monitoring of 30 mer DNA hybridization in phosphate-buffered saline solution, pH 7.4. Two other devices were used to simultaneous test complementary (CM30) and mismatched (MM30) DNA buffer solutions. Selective response to addition of complementary DNA is observed and almost no change upon addition of phosphate-buffered saline solution or mismatched DNA [29]. Copyright © 2006 American Chemical Society.

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Real-time DNA biosensors to detect cystic fibrosis genes have been demonstrated with sensitivities up to the femtomolar range [31]. These bio-FETs hinge on silicon nanowire (SiNWs) channels functionalized with peptide nucleic acid receptors that are complementary to the wild type of cystic fibrosis genes. Tailoring SiNWs is relatively easy due to the well-known modification of silicon oxide surfaces as opposed to graphene or carbon complex surfaces. Although a nonspecific interaction of negatively charged oligonucleotides with NW sensors was observed, specific detection was possible by analyzing the magnitude of the conductance change following introduction of wild vs mutant DNA sample solutions. After detection, the conductance was reversible to its original state upon addition of DNA-free solution.

Highly selective and sensitive real-time protein detection was reported with SiNW bio-FETs [25]. Nanotube/nanowire semiconductor channels offer higher surface area than planar devices and increase sensitivity to the point that single-molecule detection is possible. Sensing of calcium ions (Ca^{2+}) was possible by immobilizing calmodulin onto SiNW surfaces. A drop in conductance upon addition of 25 μM Ca^{2+} solution was observed with a subsequent recovery in conductance when the device was rinsed with a Ca^{2+} free solution, **Figure 5**. Control experiments with unmodified SiNWs did not show a change in conductance when Ca^{2+} ions were added.

SiNW arrays modified with antibodies for influenza A or adenovirus displayed selective conductance change corresponding to single-virus binding/unbinding of influenza A and paramyxovirus [32]. Many bio-FETs can be built in parallel for multiplexed biodetection.

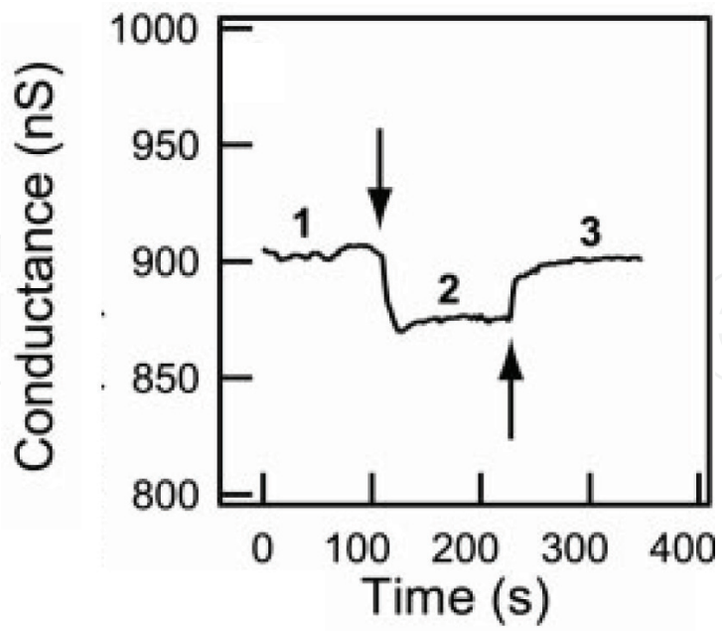

Figure 5. Real-time detection of calcium ions. Plot of conductance vs time of a bio-FET channel of calmodulin-terminated silicon nanowire where regions 1 and 3 correspond to the operation under pure buffer solution and region 2 corresponds the operation under 25 μM Ca^{2+} solution [25]. Copyright © 2001 by the American Association for the Advancement of Science.

No purification of virus samples was required in these measurements. **Figure 6** shows the conductance vs time recorded simultaneously for influenza A (nanowire 1) and adenovirus group III (nanowire 2) sensors set in proximity with a microfluidic system that allows the sequential flow of 1–4 adenovirus, influenza A, pure solution, and 1:1 mix solution of adenovirus and influenza A, respectively. Adenovirus is negatively charged in the solution resulting in a positive conductance change of nanowire 2 with an on-time duration of ca 16 s. Negative conductance change of similar duration was observed when influenza A was introduced and binded to nanowire 1. Bottom arrows in **Figure 6** indicate singularities in which the proximity of adenovirus in device 1 resulted in a short-lived positive change of conductance ca 0.4 s, and similarly, the proximity of influenza A resulted in a short-lived negative change of conductance in device 2. The excellent binding selectivity, single viral particle sensitivity and selective multiplexed detection enables rapid identification of viral samples as required for robust medical solutions, fundamental virology, and drug discovery.

Label-free amplified biodetection has been reported with floating-gate transistor architectures, **Figure 7** (left), in which the gate voltage is indirectly applied via a secondary electrolyte [33]. In such structures, the semiconductor does not need to be modified with selective bioreceptors and is not in direct contact with the analytes therefore the semiconductor can be selected on its electronic performance and doping mechanism (field-effect vs electrochemical) and not on the ease of its chemical modification or robustness in electrolytes, thereby reducing fabrication complexity. Bioreceptors are set on one part of the floating gate coupled with a secondary electrolyte compartment and entirely separated

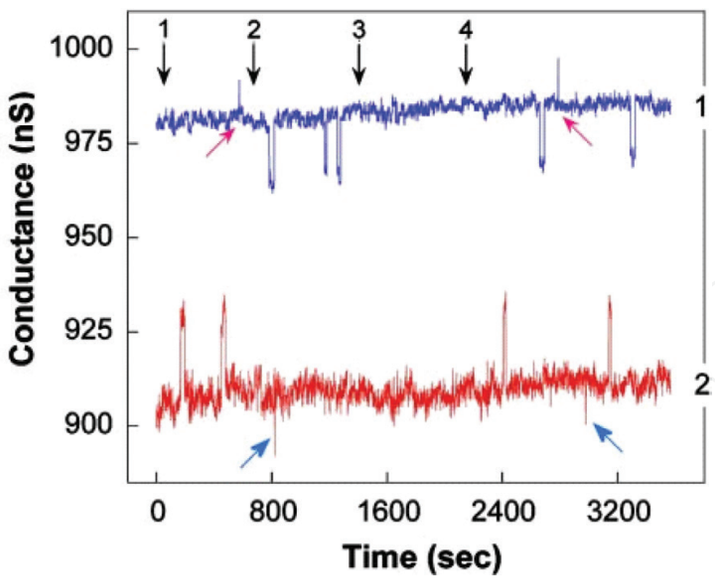

Figure 6. Conductance vs time recorded simultaneously from two silicon nanowire sensors, nanowire 1 was modified with influenza A antibody (top) and nanowire 2 was modified with adenovirus group III antibody (bottom). Arrows 1–4 correspond to the introduction of (1) adenovirus, (2) influenza A, (3) pure buffer, and (4) 1:1 mixture solution of adenovirus and influenza A. Bottom arrows highlight short-duration conductance changes corresponding to nonspecific diffusion of viral particles. Solutions made by 40 viral particles per μl in phosphate buffer 10 μM, pH 6.0 [32]. Copyright © 2014 National Academy of Sciences.

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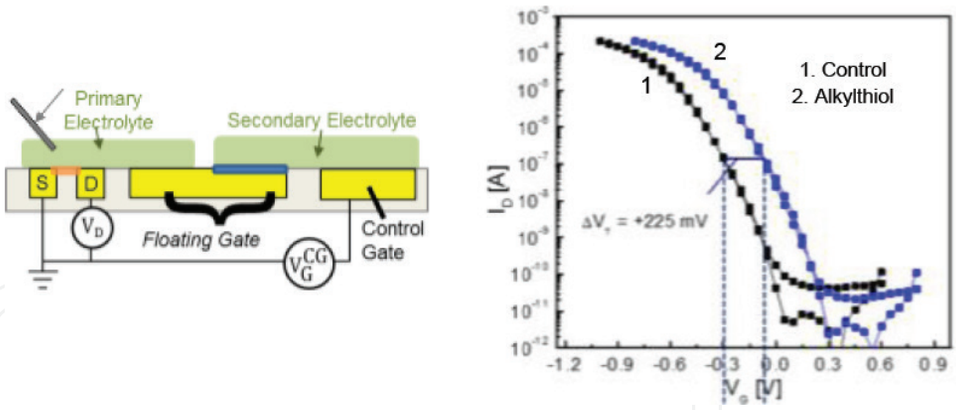

Figure 7. Device architecture of the floating-gate transistor structure (left) and transistor transfer characteristics (right), drain-source current (I_{ds}) vs gate voltage (V_{gs}), of control device, curve 1 and floating-gate functionalized with alkylthiol derivatives right-shifted 225 mV from the control device, curve 2 [33]. Copyright © 2015 American Chemical Society.

from the primary electrolyte [34–36]. The bioreceptor-free gate electrode is coupled with the primary electrolyte. Well established self-assembled monolayer chemistry on metal electrodes can be employed to tune the effective potential of the floating gate for bioelectric signal amplification [37]. Geometric considerations of the gate electrode play a crucial role in the device operation [38, 39]. Self-assembled monolayers of alkylthiol derivatives on the floating gate had a measurable voltage shift (ca. 200 mV) on the transistor electric characteristics, **Figure 7** (right).

Label-free DNA hybridization has been sensed by a floating-gate transistor based on poly(3-hexylthiophene) and an ion gel electrolytes [36]. The DNA is hybridized at the floating gate resulting in a shift in the threshold voltage of the transistor with a relative magnitude proportional to the DNA mismatch. The response of DNA with three mismatched base pairs was undistinguishable from a fully random DNA sequence.

3. Organic electrochemical transistor biosensors

Organic electrochemical transistors (OECTs) are emerging as a promising platform for amplified biodetection with enhanced sensitivity and low voltage (<1 V) operation [33, 40, 41]. OECTs typically consist of a polymer semiconductor included between source and drain electrodes and coupled with a gate electrode through an electrolyte. Polymer semiconductor materials are based on pi-conjugated carbon and hydrogen backbone structures that are relatively soft, flexible, and permeable to ionic species. Polymers can be soluble in organic solvents and be printed on flexible substrates. The gate electrode and the semiconductor channel can be approached as the electrodes of a conventional electrochemical sensor (two electrodes) assigning the transistor channel together with drain and source electrodes as the working electrode and the gate electrode as the counter electrode. By modifying the channel, electrolyte or gate electrode with biochemical receptors, chemical binding events can result in large changes in conductivity in the semiconducting channel. This effect can be the basis for bioelectric signal amplification [28, 34, 42, 43].

To mitigate the reproducibility and sensibility issues related with electrochemical biosensors, a simplified electrochemical transistor architecture can be employed in which the use of high surface area activated carbon electrodes results in sub-1 V operation and renders unnecessary presence of a reference electrode to control the applied potentials [38, 44].

Early OECTs consisted in Au microelectrodes drain-source and gate electrodes and polypyrrole channels in aqueous electrolyte media, i.e., CH_3CN/0.1 M [n-Bu_4N]ClO_4 displaying typical transistor p-type characteristics [45]. When the gate is held at negative voltages, polypyrrole is dedoped and the device is switched off. When the gate voltage is positive, polypyrrole is oxidized resulting in an increase in the channel current, the device is switched on.

Nowadays, polystyrenesulfonate-doped poly(3,4-ethylenedioxythiophene) (PEDOT:PSS) is the most commonly employed channel material in OECTs. PEDOT:PSS is intrinsically a p-type conducting polymer, stable in water, and nontoxic.

Enzymatic glucose OECT biosensors have been reported [46–48]. Hydrophilic ionic liquids like triisobutyl-(methyl)-phosphonium tosylate ([$P_{1,4,4,4}$][Tos]) are particularly interesting as effective enzyme immobilization medium in biological environments [49]. Sensing experiments were carried out by placing enzyme-glucose oxidase (GOx, 500 units per mL) and ferrocene mediator [bis(η-5-cyclopentadienyl)iron] (Fc, 10 mM) in 1.43 mL of [$P_{1,4,4,4}$][Tos] and 50 mL glucose-PBS solution.

Glucose is oxidized by the application of a gate voltage while the enzyme (GOx) is reduced. In the back cycle, oxidation of GOx is coupled with the conversion of Fc to ferricenium ion (Fc^+) which shuttles electrons to the gate electrode (**Figure 8a**) and cations into PEDOT:PSS. PEDOT:PSS is dedoped by metal cations (M^+) (**Figure 8b**) decreasing the drain-source current as a function of glucose concentration. The measured sensitivity of such devices is in the range of 10^{-7}–10^{-2} M [49].

OECTs hormone biosensors have been reported by incorporating Nafion, a material that has high specific affinity for epinephrine hormone molecules, on top of the gate electrode [50]. The sensitivity was improved by introducing gate electrodes modified with SWNTs and graphene flakes which enhanced electrocatalytic activity of the gate electrode compared to that with platinum gates thus improving the detection limit of the device. Sensitivities up to 0.1 nM were obtained.

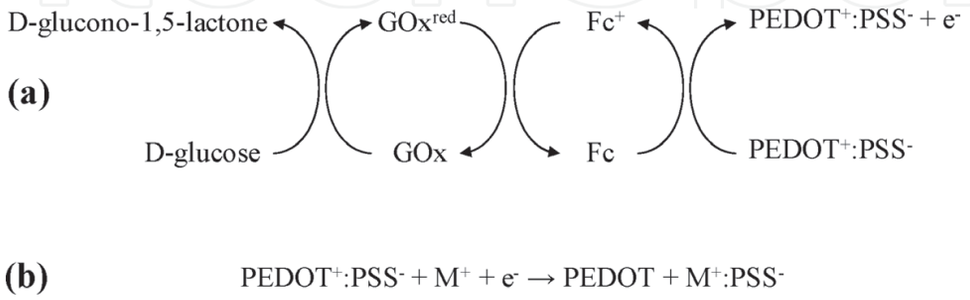

Figure 8. Reactions at (a) the gate electrode and (b) the channel of the OECT [49]. Copyright © 2010, Royal Society of Chemistry.

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Label-free, flexible DNA OECT sensors have been demonstrated in flexible microfluidic systems [51]. Single-stranded DNA (ssDNA) probes were immobilized on gold gate electrodes and poly(dimethylsiloxane) (PDMS) microfluidic channels were fixed on top of prepatterned PEDOT:PSS channels (**Figure 9a**) set in contact with gold drain and source electrodes. The device was able to bend on both sides (**Figure 9b**) and its electric characteristics remained consistent before and after bending (**Figure 9c**). Transient curves indicate that the drain-source current reaches a stable plateau value after several seconds of applied gate voltages (**Figure 9d**) revealing ion permeation into the transistor channel [52]. Further investigations revealed that the maximum bending strain was about 5% with negligible effect on the PEDOT:PSS film conductance.

Biocompatible materials have been investigated due to the rapid development of OECTs and their potential use in biological applications. Skin itself can be utilized as a nonconventional gate electrode to sense heart beatings, **Figure 10a**. Campana et al. demonstrated transparent transistors fabricated on biodegradable poly(lactic-co-glycolic acid) substrates to sense heart beatings [53]. Traditional electrocardiograms utilize Ag/AgCl to establish a Faradaic contact between the skin and the electrodes and to sense small voltage

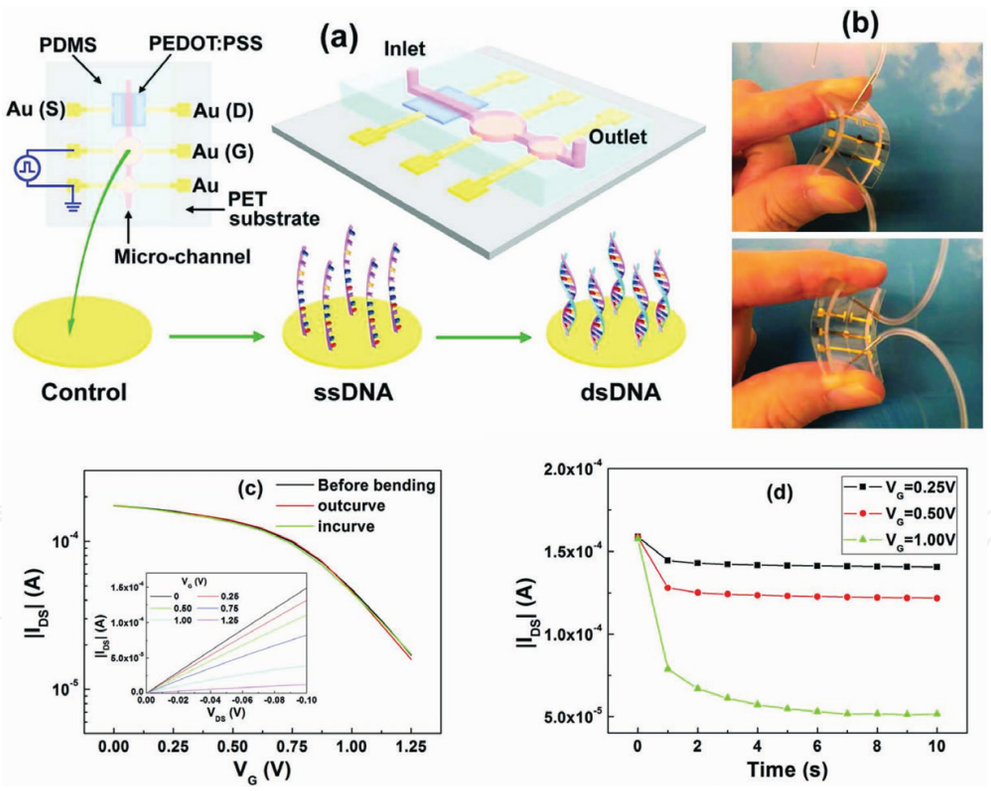

Figure 9. (a) Schematic diagram of the OECT integrated with a flexible microfluidic system and gate gold electrodes before (control) and after DNA modification and DNA hybridization. (b) Photographs of a device bent on both sides. (c) Transfer characteristics (I_{ds} vs V_{gs}) of the OECT measured before and after bending it on both sides, inset shows the output characteristics (I_{ds} vs V_{gs}). (d) Time-dependent channel current of the OECT measured after applying different gate voltages. The drain-source voltage (V_{ds}) in the transfer and time-dependent characteristics was −0.1 V [51]. Copyright © 2011 WILEY-VCH Verlag GmbH & Co. KGaA, Weinheim.

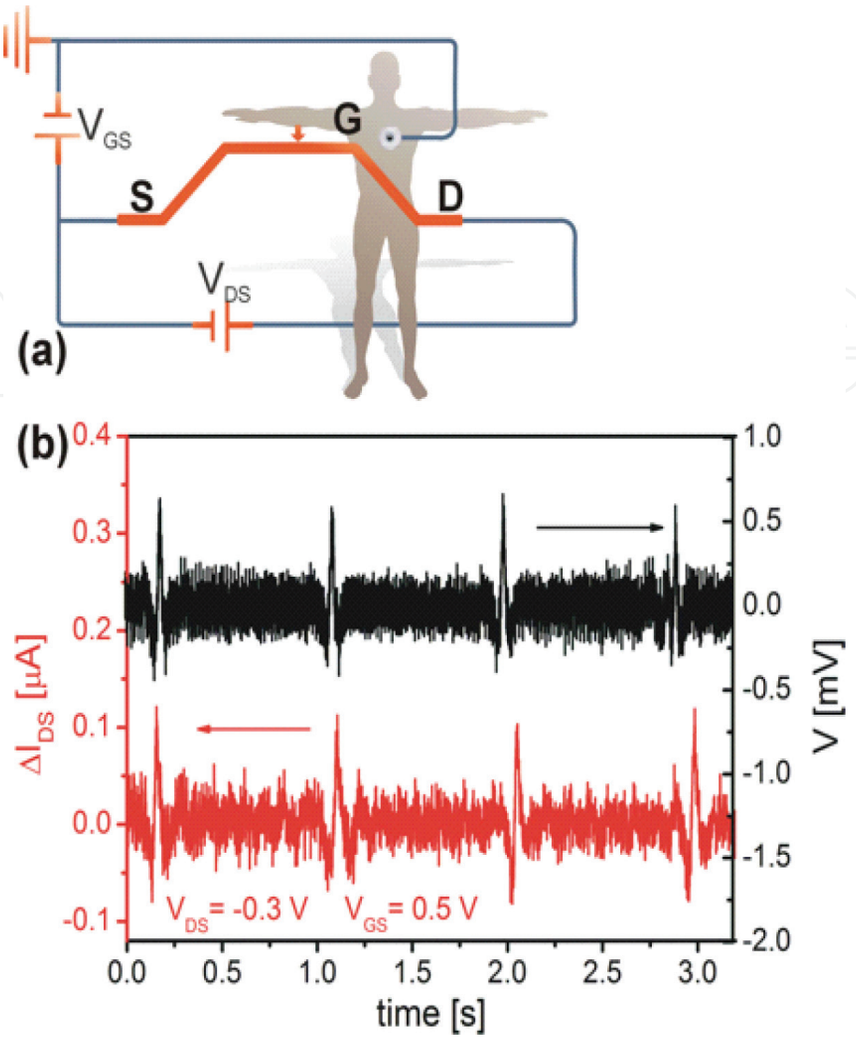

Figure 10. ECG recording with a bioresorbable OECT operated in direct contact with the skin. (a) Wiring diagram of the experiment. (b) Measured drain-source current (bottom, left axis) as obtained during transistor recording (V_{gs} = 0.5 V, V_{ds} = −0.3 V) and comparison with a normal electrocardiogram potentiometric recording with standard disposable leads (top, right axis) [53]. Copyright © 2014 WILEY VCH Verlag GmbH & Co. KGaA, Weinheim.

perturbations that exist on the skin per heart beatings [54]. To reduce the skin's impedance, an electroconductive gel was employed between the skin and PEDOT:PSS channels. The OECTs recorded heartbeats with I_{ds} amplitude of ca 0.1 μA equivalent to measuring signals at the gate with amplitudes of ca 50 μV whereas traditional electrocardiograms monitor spikes at ca 500 μV, **Figure 10b**.

Wearable electrochemical transistors as a platform for real-time detection of biomarkers in external biological fluids was demonstrated with a simple device structure [55]. Drain, source, and gate PEDOT:PSS electrodes were screen printed on textile fabrics. Adrenaline, dopamine, and ascorbic acid in artificial sweet were sensed in the order of tenths of μM concentrations. The device operation was stable despite several hand-washing cycles and deformations.

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Flexible lactate sensors with ionogel solid-state electrolytes have been developed [56]. The concentration of lactate in blood can indicate circulatory effectiveness of anaerobic metabolism. The ionogel was prepared by dissolving ferrocene mediator (Fc) in hydrophilic ionic liquid 1-ethyl-3-methylimidazolium ethylsulfate [C_2mIm][$EtSO_4$] followed by mixing it with monomer N-isopropylacrylamide (NIPAAm) cross-linker N,N'-methylenebis(acrylamide) (MBAAm) and photoinitiator (dimethoxyphenyl)acetophenone DMPA. Under application of a gate voltage, lactic acid was oxidized to pyruvate and Fc was converted to ferricenium ion (Fc^+). Fc^+ transduce electrons to the gate electrode and the PEDOT:PSS channel is dedoped with cations from the solution which leads to a decrease in the drain-source current.

A disposable alcohol breath sensor, to estimate the content of alcohol in the blood, was demonstrated using organic electrochemical transistors [57]. Alcohol dehydrogenase and nicotinamide adenine dinucleotide (NAD^+) enzymes were immobilized on a collagen-based gel. PEDOT:PSS was employed as drain, source, and gate electrodes. When NAD^+ is in contact with ethanol and alcohol dehydrogenase, it can get reduced to NADH and oxidized back to NAD^+, freeing two electrodes that can change the conductivity state of PEDOT:PSS.

4. Conclusion

In this chapter, we reviewed emerging concepts of biosensors based on transistor structures with relevance in the field of easy-to-use, portable, and user-friendly devices for preventive personalized medical applications and point-of-care testing. Soaring healthcare costs, trips to the hospital and the stress and pain related to draw blood for testing represent major inconveniences in patients that require being examined regularly, i.e., insulin-dependent. We believe that the staggering progress done in biosensors based on field-effect transistors and organic electrochemical transistors represents a major opportunity to empower people globally with medical information in a timely and cost-effective way.

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