Highly Sensitive and Wearable In$_2$O$_3$ Nanoribbon Transistor Biosensors with Integrated On-Chip Gate for Glucose Monitoring in Body Fluids

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*Supporting Information

ABSTRACT: Nanoribbon- and nanowire-based field-effect transistor (FET) biosensors have stimulated a lot of interest. However, most FET biosensors were achieved by using bulky Ag/AgCl electrodes or metal wire gates, which have prevented the biosensors from becoming truly wearable. Here, we demonstrate highly sensitive and conformal In$_2$O$_3$ nanoribbon FET biosensors with a fully integrated on-chip gold side gate, which have been laminated onto various surfaces, such as artificial arms and watches, and have enabled glucose detection in various body fluids, such as sweat and saliva. The shadow-mask-fabricated devices show good electrical performance with gate voltage applied using a gold side gate electrode and through an aqueous electrolyte. The resulting transistors show mobilities of $\sim 22 \text{ cm}^2\text{ V}^{-1}\text{ s}^{-1}$ in 0.1× phosphate-buffered saline, a high on–off ratio (10$^5$), and good mechanical robustness. With the electrodes functionalized with glucose oxidase, chitosan, and single-walled carbon nanotubes, the glucose sensors show a very wide detection range spanning at least 5 orders of magnitude and a detection limit down to 10 nM. Therefore, our high-performance In$_2$O$_3$ nanoribbon sensing platform has great potential to work as indispensable components for wearable healthcare electronics.

KEYWORDS: wearable biosensor, gold side gate, glucose sensor, indium oxide semiconductor, field-effect transistor, shadow-mask fabrication

Wearable biosensors are smart electronic devices that can be worn on the body as implants or accessories. Recent advances in microelectronics, telecommunications, and sensor manufacturing have opened up possibilities for using wearable biosensors to continuously monitor an individual’s body status without interrupting or limiting the user’s motions. However, while many commercially available wearable electronics can track users’ physical activities, devices that can provide an insightful view of user’s health status at the molecular level need more development. On the other hand, although some commercial hand-held analyzers enable glucose or lactate detection, most of these devices rely on blood samples. Neither finger-prick nor invasive sensors (such as a needle embedded under the skin) are desired for wearable biomedical applications. Continuous analyte monitoring, a key advantage offered by wearable biosensors, has great potential in...
many cases. For example, optimum diabetes management needs regular glucose monitoring, and a trend of glucose level is more meaningful than an accurate data point.\textsuperscript{10} Besides glucose monitoring, real-time detection of some pathogens in body fluids can warn of the possible onset of certain diseases.\textsuperscript{11}

Although blood is by far the most understood sample for diagnosis, other biological fluids such as sweat, tears, and saliva also contain tremendous biochemical analytes that can provide valuable information and are more readily accessible compared to blood.\textsuperscript{10,12} Recent studies suggest a diagnosis system based on the glucose concentration in body fluids to estimate blood glucose levels.\textsuperscript{13–15} However, many challenges still exist for the accurate glucose sensing in body fluids.\textsuperscript{16,17} For example, the glucose levels in body fluids are much lower than that in blood.\textsuperscript{18} The sensing results can be affected by ambient temperature changes, mechanical deformation caused by body motion, and the sample collection procedure.

Among various types of sensors (e.g., optical, piezoelectric, and electrochemical sensors), electrochemical sensors are the most promising candidate for wearable technology owing to their high performance, portability, simplicity, and low cost.\textsuperscript{19–24} Considering the demands of wearable biosensors, the selection of the sensing platform is critical to high sensitivity and reproducibility, real-time detection, and simple integration with wearable environments (e.g., human skin, tooth, and eye).\textsuperscript{11,25} Nanobiosensors based on indium oxide (In\textsubscript{2}O\textsubscript{3}) field-effect transistors (FETs) are well suited for wearable biosensor applications because of the quick response time for real-time and continuous monitoring, large detectable concentration range, high sensitivity, high uniformity for reliable sensing, and the capability to integrate with other microfluidic and electronic functional groups.\textsuperscript{26–30} Furthermore, the exposed semiconductor channel regions can be modified with various functional groups or receptors easily and thus make the In\textsubscript{2}O\textsubscript{3} nanobiosensors suitable for multiplexed sensing.

The current FET-based biosensor platform is usually composed of individual sensors with an external Ag/AgCl solution gate electrode to set the operational point of the sensors to the optimal detection mode. In order to build a wearable biosensor platform, a stand-alone sensor array is desired. The Ag/AgCl electrode is commonly used as the...
Here, we demonstrate highly sensitive and conformal In$_2$O$_3$ nanoribbon FET biosensors with a fully integrated on-chip gold side gate, which have been laminated onto various surfaces, such as artificial arms and watches, and have enabled glucose detection in various body fluids, such as sweat and saliva. The devices are fabricated through two shadow masks. First, a shadow mask is used to define the sputter-coating of In$_2$O$_3$ nanoribbons, and the second shadow mask is used for metal deposition of the source, drain, and side gate. The source and drain electrodes are modified with the enzyme glucose oxidase (GOx), biocompatible polymer chitosan, and single-walled carbon nanotubes (SWCNTs) using inkjet printing. The gold side gate In$_2$O$_3$ FETs show good electrical performance on highly flexible substrates. The optimized glucose sensors show a very wide detection range spanning at least 5 orders of magnitude and detection limits down to 10 nM. The noninvasive glucose detections in human body fluids, such as tears and sweat, and the sensing on artificial skin and eye replicas are demonstrated. Therefore, this type of device is a highly sensitive platform for not only glucose detections but also many other types of sensing applications.

RESULTS AND DISCUSSION

The In$_2$O$_3$ nanoribbon devices were fabricated following our previous reported shadow-mask fabrication technique, however, this time we have added side gate patterns to the source/drain shadow mask and have used a 5 μm ultraflexible PET substrate. Figure 1a illustrates the scheme for fabricating flexible In$_2$O$_3$ macrowire electronics on PET substrates. First, a PET substrate was attached to the first shadow mask using antistat tape. Then we used radio frequency (RF) sputtering to deposit 16 nm thick In$_2$O$_3$ nanoribbons through the openings on the shadow mask. The second shadow mask was then laminated onto the PET substrate for the following metal deposition. After using a single mask to define the source, drain, and side gate electrodes, the as-made biosensor foil was peeled off from the shadow mask for further electrical characterization. While most previous glucose-sensing studies used electrochemical sensors with large working electrodes by functionalizing the electrodes with drop casting, here, we have developed an inkjet printing technique to functionalize our FET In$_2$O$_3$ glucose biosensors, as shown in Figure 1b. Due to the small dimensions (∼25 μm × 500 μm) of our nanoribbon biosensors, the traditional drop-cast deposition method causes the whole active sensing area to be covered by the chitosan film. In order to keep the channel area exposed, we employed a SonoPlot printer with a 50 μm glass nozzle to print the chitosan ink only on the source and drain pads. The ink was made of chitosan, single-walled carbon nanotubes, and glucose oxidase, and the preparation of the ink is described in detail in the Experimental Methods section. Figure 1c shows a photograph of an as-fabricated In$_2$O$_3$ biosensor foil with a size of 5 cm × 5 cm. An optical image of a group of In$_2$O$_3$ biosensors and two gold side gate electrodes is shown in Figure 1d. Figure 1e shows a scanning electron microscope (SEM) image of the channel region and the gold side gate of a biosensor. To further characterize the In$_2$O$_3$ nanoribbons, we used atomic force microscopy (AFM) and X-ray diffraction (XRD) on samples deposited on a polyethylene terephthalate (PET) substrate (Figure S1). The AFM images show that the nanoribbons are solid and have clear edges. The height profile shows the thickness of In$_2$O$_3$ nanoribbons is ~20 nm. The XRD pattern shown presents only PET peaks, indicating the In$_2$O$_3$ is amorphous. Figure 1f shows the fabricated In$_2$O$_3$ nanoribbon FET foil was conformably laminated onto an artificial human hand, demonstrating the bendability and wearability of the In$_2$O$_3$ nanoribbon biosensors. Figure 1g exhibits the biosensor foil rolled up with a radius of curvature of ~1 mm. The flexible biosensor can be further attached onto the back casing of a watch (Figure 1h), showing the concept that such In$_2$O$_3$ transistor biosensors can be integrated with smart watches in the future. We believe that flexible, lab-on-a-chip, and conformal In$_2$O$_3$ nanoribbon electronics are highly advantageous for wearable biosensor applications.

In many previous studies, a Ag/AgCl electrode is commonly used as the reference electrode in electrochemical measurements and biosensing applications because it can provide stable potential and can also read voltage precisely. However, the integration of the Ag/AgCl electrode onto a biosensor chip increases the steps and difficulty of fabrication. Herein, we have replaced the Ag/AgCl external electrode with a gold side gate to supply gate bias to the devices. There are two gold side gate electrodes in a group of four In$_2$O$_3$ FET devices. The one in the middle will replace the Ag/AgCl external liquid gate to supply gate voltage, and the other one at the rear can be used to monitor changes in potential on the devices. We first compared the device performance with gate voltage applied by the external Ag/AgCl electrode and the on-chip gold side electrode. Here, all the measurements were done when the device active area was immersed into a micro-well filled with 300 μL of electrolyte solution (0.1x phosphate buffered saline (PBS)). Figure 2a and b show family curves of drain current–gate voltage ($I_{DS}$–$V_{GS}$) and drain current–drain voltage ($I_{DS}$–$V_{DS}$) when the gate voltage was biased through a Ag/AgCl electrode. The schematic diagram of the measurement setup is shown in the inset of Figure 2a. The performance of the gold side gate controlled In$_2$O$_3$ FET is presented in Figure 2c ($I_{DS}$–$V_{GS}$) and d ($I_{DS}$–$V_{DS}$). The output and transfer curves of the FET devices demonstrate that the In$_2$O$_3$ nanoribbon devices can work properly under gate bias supplied by the gold side gate. The output characteristics of the FET devices demonstrated Ohmic behavior with a good linear regime in the “on” state, and the drain current got saturated when the bias increased further. All the curves in Figure 2b and d passing through the origin point indicate the minimal contribution of the gate leakage current to the drain current. The field-effect mobility of the
In$_2$O$_3$ FET is extracted to be $22.34 \pm 1.44$ cm$^2$ V$^{-1}$ s$^{-1}$ using the following equation:

$$g_m = \frac{dI_D}{dV_G} = \frac{W}{L} C_{DL} \mu_{FE} V_D$$  \hspace{1cm} (1)$$

where $W$ is the channel width, $L$ is the channel length, and $C_{DL}$ is the electrical double layer capacitance per unit area in 0.1 M ionic strength aqueous solution reported previously (25.52 $\mu$F cm$^{-2}$). The maximum transconductance of 5.69 $\mu$S was obtained at a drain voltage of 0.2 V and a gate voltage of 0.527 V (Figure S2 in the Supporting Information). To further confirm the gate control of the on-chip side gate electrode, we used one electrode as the gate bias supplier and another one as a reference electrode to monitor the actual change of potential on the devices, as the scheme shows in the inset of Figure 2e. In Figure 2e, we plotted the reference voltage ($V_{REF}$) against the gold side gate voltage ($V_{GS}$) with different distances between those two electrodes, 150, 750, and 1350 $\mu$m, respectively. It shows that $V_{REF}$ is almost identical to $V_{GS}$ regardless of the distance. We further plotted drain current versus gate bias applied through the gold side gate at different distances (Figure 2f), and the curves show negligible differences. A statistical study of key electrical properties for 50 In$_2$O$_3$ nanoribbon devices comparing the gate biased through the Ag/AgCl electrode and the gold side gate was conducted. Figure S3 in the Supporting Information exhibits the device performance including mobility ($\mu$), threshold voltage ($V_{th}$), on/off ratio, and on-state current, which are very similar to each other, which implies that the gold side gate and the Ag/AgCl gate can give an analogous gating effect. From all these figures of merit, we can conclude that the on-chip gate electrode has great control over the nanoribbon transistors in an aqueous environment.

In order to characterize the flexibility of the wearable In$_2$O$_3$ FETs, bending tests were carried out. As shown in Figure 3a, we tightly wrapped our fabricated In$_2$O$_3$ foil around a cylinder. The electrical performance of the devices under tensile strain was measured. Figure 3b compares the transfer characteristics of a representative In$_2$O$_3$ FET under a relaxed state, bent with a radius of $\sim$3 mm, and after bending 100 times. (c) Mobility, (e) threshold voltage, and (g) on–off ratio of In$_2$O$_3$ FETs bent with different radii. (d) Mobility, (f) threshold voltage, and (h) on–off ratio of In$_2$O$_3$ FETs bent with a radius of $\sim$3 mm after different bending cycles.
In2O3 FETs functionalized with a gel film containing chitosan, SWCNTs, and glucose oxidase. Figure 3c, e, and f plot the mobility, the on-off ratio, and the threshold voltage averaged over nine devices bent with a radius of curvature of infinity (relaxed) and 3, 10, and 15 mm, respectively. The typical transfer curves of the devices under different bending conditions are plotted in Figure S4 in the Supporting Information. When the foil was bent with a radius of curvature of ~3 mm, a tensile strain of ~0.25% (Supporting Information, S5), parallel to the drain-to-source current direction, was applied to the In2O3 FETs. We have further plotted out the mobility as a function of strain (Figure S5 in the Supporting Information). There was no significant change of the electrical performance of the In2O3 FETs when the devices were in different bending conditions, as the mobility only showed a small variation between 22.15 ± 1.68 and 22.70 ± 1.65 cm² V⁻¹ s⁻¹, the threshold voltage only showed a variation between 0.273 ± 0.028 and 0.280 ± 0.027 V, and the logarithm on-off ratio showed a variation between 4.71 ± 0.13 and 4.84 ± 0.12.

Figure 3d, f, and h plot the mobility, the threshold voltage, and the on-off ratio of the devices after 0 (before bending), 5, 10, 50, and 100 bending cycles, respectively, and the changes in device performance were negligible as well. The mobility varied in the range of 22.98 ± 1.34 to 23.78 ± 1.87 cm² V⁻¹ s⁻¹, the threshold voltage varied between 0.273 ± 0.005 and 0.266 ± 0.016 V, and the logarithm on-off ratio varied between 4.98 ± 0.17 and 4.96 ± 0.14. On the basis of the test results, all the In2O3 nanoribbon FETs after bending tests still maintained excellent performance, confirming that our platform is reliable under mechanical deformation.

The ability to sense in a small amount of liquid is crucial to wearable sensors, because of the limited amount of body fluids at regular intervals. A polydimethylsiloxane (PDMS) stamp was adopted as a microwell to accumulate body fluids (Figure 4a). It can also serve as a passivation layer to ensure reliable sensing without disturbance introduced by electrical contact of the metal lines with the body and body fluids. A mixture of curing agent and PDMS at a ratio of 1:10 was first spin-coated onto a silicon wafer before thermally curing at 80 °C for 1 h. After punching a hole with a diameter of 3 mm, the PDMS stamp was laminated onto the biosensor substrate by van der Waals force. To guarantee the biosensor can work properly in a limited amount of liquid, we filled the PDMS microwell with 10 μL of solution and performed electrical measurements using a gold side gate electrode. Figure 4b and c show the transfer curves and output curves of the In₂O₃ FETs measured with a gold side gate in the electrolyte of ~10 μL of 0.1X PBS. The electrical performance measured in a small amount of liquid is comparable to the results shown in Figure 2c and d (measured in 300 μL of 0.1X PBS). It illustrates that our biosensing platform can efficiently work in the liquid with an amount as small as 10 μL, which is a 30-fold decrease from what we previously reported.

Figure 4 shows a schematic diagram depicting the working principle of a glucose sensor. (f) Glucose sensing results in 0.1X PBS with a gold side gate.

After ensuring that our devices has good electrical performance and ionic sensitivity with the gold side gate, the In₂O₃ nanoribbon biosensors were used to detect D-glucose. Figure 4e shows a schematic diagram depicting the working principle of the glucose determination using In₂O₃ nanoribbon biosensors. The surfaces of the source and drain electrodes were functionalized with chitosan/carbon nanotubes/glucose oxidase using inkjet printing (see Experimental Methods for details). Chitosan is chosen to work as the immobilization layer since it is a biocompatible polymeric matrix with good film-forming ability and high water permeability. Carbon nanotubes have been reported as efficient routes for increasing the sensitivity for many types of sensors, owing to their good electrocatalytic

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property and capacity for biomolecule immobilization.\textsuperscript{24,35,36} After being immobilized by the chitosan film and carbon nanotubes, the glucose oxidase enzymes accept electrons when they interact with glucose in the solution and thereafter transfer electrons to molecular oxygen, consequently producing hydrogen peroxide (H\textsubscript{2}O\textsubscript{2}). The enzymatically produced H\textsubscript{2}O\textsubscript{2} will be oxidized under a bias voltage. The reactions are as follows:

\[ \text{D-glucose} + O_2 \xrightarrow{\text{glucose oxidase}} \text{gluconic acid} + H_2O_2 \]  
\[ H_2O_2 \rightarrow O_2 + 2H^+ + 2e^- \]  

(2) (3)

The generation of H\textsuperscript{+} depends on the concentration of glucose. Decreasing of the pH leads to protonation of the OH groups on the In\textsubscript{2}O\textsubscript{3} surface and results in changes in the local FET electric field, ultimately causing changes in the conductance and current. Figure 4f shows the continuous monitoring of the sensing signal in response to different glucose concentrations. The channel current increases with the additions of glucose and shows the detection limit of about 10 nM (~2.2% of the baseline current). Our glucose sensor can detect glucose in the concentration range between 10 nM and 1 mM, covering typical glucose concentrations in human body fluids. These glucose concentrations correspond to typical sweat glucose concentrations of both diabetes patients and healthy people.\textsuperscript{10} The detection limit we obtained here is much lower than a typical electrochemical amperometric glucose sensor.\textsuperscript{7,8} We also performed control experiments on sensors without glucose oxidase, and as observed, those sensors did not respond to glucose (Figure S7 in the Supporting Information).

Wearable In\textsubscript{2}O\textsubscript{3} nanoribbon glucose sensors are further used for human body fluid analysis. The glucose concentration is much lower in tears, sweat, and saliva than in blood. While normal blood glucose levels range between 70 mg/dL (3.9 mM) and 140 mg/dL (7.8 mM) or higher, by contrast, tear glucose levels are on the order of 0.1–0.6 mM,\textsuperscript{37–39} sweat glucose has been reported at 5 to 20 mg/dL (0.277–1.11 mM),\textsuperscript{18} and saliva glucose concentrations are around 0.51–2.32 mg/dL (28.3 \textmu M to 0.129 mM).\textsuperscript{40–42} Figure 5a, b, and c show the representative current responses of the glucose in artificial human tears, artificial human sweat, and saliva, respectively. The details about the preparation of the body fluids are described in the Experimental Methods section. Initially, in Figure 5a, the devices were submerged in 0.1X PBS to obtain the baseline current. After changing the electrolyte from 0.1X PBS to artificial tears at 150 s, the sensing signal bumped up a little bit, which is due to the pH difference between artificial tears and 0.1X PBS. Noise levels of the glucose sensing in artificial tears were higher than the results in PBS when comparing the inset figures in Figure 4f and Figure 5a. This high noise level comes from the weaker buffer capability of the artificial tears and results in a lower signal-to-noise ratio, which consequently affects the detection limit. We extracted the relationship between the glucose concentration and the saturated current response from the real-time sensing data in PBS solution and plotted it in Figure 5d. For comparison, the sensing results in artificial tears, sweat, and saliva are also plotted. The high agreement between the data with PBS and the data with artificial tears is a good indicator that the signals from both media are attributed to mainly glucose instead of other nonspecific proteins. In the cases of artificial sweat and saliva, even though the sensing signals are slightly lower than the responses from PBS, which may be due to their different ionic strengths and complex ingredients, our sensors can differentiate a glucose concentration as low as 0.1 \textmu M. This sensitivity is sufficient to detect glucose in both sweat and saliva.

The In\textsubscript{2}O\textsubscript{3} biosensors can be comfortably attached onto an artificial eyeball and an artificial arm (Figure 6a). To ensure the on-body sensing ability, we imitated the data collection on an artificial eyeball with the biosensor facing out. Figure 6b shows the \textit{ex situ} glucose-sensing results using artificial tears. After using indium wires to connect the bonding pads to our measurement unit, we constantly flowed artificial tears through the sensing area, as shown in the inset of Figure 6b. After obtaining a stable baseline current, we sequentially flowed artificial tears spiked with 0.01, 0.1, 1, 10, 100, and 1000 \textmu M glucose, respectively. Overall, it demonstrates that our wearable glucose-sensing platform has the potential to work as contact lenses with embedded sensors for monitoring the tear glucose level. Similarly, we performed glucose sensing on an artificial arm, but with the sensor facing down on the skin. As the sensing results shown in Figure 6c show, our In\textsubscript{2}O\textsubscript{3} biosensor can work as a sweat patch for glucose monitoring. To further confirm that our sensing platform can be utilized as a wearable sweat analyzer, we collected sweat samples from a human subject’s forehead during exercise. After we spiked the real sweat with different concentrations of glucose, glucose sensing was performed with the prepared samples. Figure 6d shows the sensing results with real sweat. After we added real sweat to replace the PBS solution, the sensing signal shows a large increase, which is because the original real sweat sample has a different pH and contains glucose before spiking. Good sensitivity was observed ranging from 0.1 \textmu M to 1 mM, indicating that our sensing platform is a great candidate for wearable sweat analysis. We also measured sweat glucose levels before and after a meal for a healthy person. The before- and after-meal sweat sample were collected 30 min before and 30 min after a glucose beverage intake. The sensing results are shown in Figure 6e; the inset figure shows the device transfer
curve measured using before-meal and after-meal sweat samples as electrolyte. The subject’s blood sugar level before and after a meal is also recorded using a commercial glucose meter, giving readings of 79 and 118 mg/dL, respectively.

We previously demonstrated that our In$_2$O$_3$ biosensors showed a very stable performance when they were kept in aqueous solutions. However, the proteins such as glucose oxidase may not be as robust as the sensors after a long time. The device was measured under ambient conditions. Recently, great progress has been made using liquid metal-based reaction to produce 2D semiconductors. Because of their high charge-carrier mobility and outstanding mechanical properties, 2D materials are very promising for next-generation wearable electronics. Wearable biosensors using single-crystalline, sub-nanometer layers of 2D materials, such as In$_2$O$_3$, Ga$_2$O$_3$, SnO$_2$, ZnO, and InGaZnO, can be explored in the future.

CONCLUSIONS

In summary, the In$_2$O$_3$ FET-based wearable biosensors with on-chip gold side gate electrodes can be used for highly sensitive detection of glucose with a detection limit down to 10 nM. The all-on-a-chip device structure can be incorporated into a straightforward two-step shadow-mask fabrication. The gold side gate electrodes show a stable and efficient gating effect on In$_2$O$_3$ FETs on flexible substrates. Mobilities in 0.1X PBS of $\sim$22 cm$^2$ V$^{-1}$ s$^{-1}$ and on–off ratios of more than 10$^5$ were achieved. The noninvasive glucose detection in human body fluids, such as tears and sweat, was also demonstrated. We further investigated glucose sensing on an eyeball replica and on an artificial hand. Lastly, we demonstrated that our glucose sensor can work in real human sweat and can distinguish glucose levels before and after a meal. Given the facile and highly scalable fabrication process, low driving voltage, and reliable sensing behavior even when deformed, this sensing platform is promising for continuous personalized health monitoring, for the food industry, and for environmental monitoring.

EXPERIMENTAL METHODS

Fabrication Process. A PET substrate was first cleaned with acetone and isopropyl alcohol and then went through ultraviolet treatment before the fabrication process. After the cleaning process, the first shadow mask was attached to the PET substrate to define the channel area. Then In$_2$O$_3$ nanoribbons were deposited by RF sputtering (Denton Discovery 550 sputtering system). By simply detaching the shadow mask, we got well-patterned nanoribbons. The source, drain, and gold side electrodes were then defined by the second shadow mask, followed by electron beam evaporation of 1 nm Ti and 50 nm Au. After deposition, the shadow mask was removed.

Characterization. Optical microscopy images were taken with an Olympus microscope. The SEM images were taken with a Hitachi S-4800 field emission scanning electron microscope. Electrical characteristics and sensing results were measured with an Agilent 1500B semiconductor analyzer.

Device Functionalization. Chitosan powder (1 wt %) was first dissolved in a 2 wt % acetic acid aqueous solution. Next, the chitosan solution was mixed with single-walled carbon nanotubes (2 mg mL$^{-1}$ in 1X PBS) using ultrasonication for over 30 min. The chitosan/SWCNT solution was mixed with a glucose oxidase solution (10 mg mL$^{-1}$ in 1X PBS) in the volume ratio of 2:1. The mixed solution was then inkjet printed onto the source and drain electrodes and dried under ambient conditions.

Human Body Fluid Samples. Artificial human sweat solution was bought from Walgreens. Artificial human sweat was prepared by mixing 22 mM urea, 5.5 mM lactic acid, 3 mM NH$_4^+$, 100 mM Na$^+$, 10 mM K$^+$, 0.4 mM Ca$^{2+}$, 50 $\mu$M Mg$^{2+}$, and 25 $\mu$M uric acid with varying

Figure 6. Off-body glucose sensing. (a) Photographs of the In$_2$O$_3$ biosensors attached onto an eyeball replica and an artificial arm. (b) Real-time glucose-sensing results on an artificial eyeball. (c) Real-time glucose-sensing results on an artificial arm. (d) Real-time glucose sensing with real sweat collected from human subjects. (e) Real-time glucose-sensing results on real sweat collected before and after glucose beverage intake.
glucose concentrations.” Real sweat samples were collected from humans by scratching their foreheads with microtubes.

**ASSOCIATED CONTENT**

Supporting Information

The Supporting Information is available free of charge on the ACS Publications website at DOI: 10.1021/acsnano.7b06823.

AFM and XRD characterization; mobility calculation; statistical study; bending test; tensile strain calculation; pH sensing; glucose-sensing control experiment; stability tests (PDF)

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**Notes**

The authors declare no competing financial interest.

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