Charge-Coupling Extended-Gate Amorphous-InGaZnO-Based Thin-Film Transistor for Use as Ultrasensitive Biosensor

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In this study, ultrasensitive pH sensing is achieved by charge-coupling with an amorphous(a)-InGaZnO-based thin film transistor (TFT) for use as an ultrasensitive biosensor. To enhance the detection sensitivity using the charge-coupling circuit, an extended-gate a-InGaZnO-based TFT sensor is employed, in which the size of the extended electrode with an oxide film is fixed but the channel size is controllable. The pH sensitivity was enhanced to 313 mV/pH at 25 °C, above the ideal potential limit, by decreasing the channel size of the charge-coupling extended-gate TFT, although an insulated-gate TFT showed a standard pH response near the ideal Nernstian potential (59.1 mV/pH at 25 °C) because of the non-charge-coupling gate structure. This is because ionic charges at an extended electrode with an oxide film, which are unrelated to the Nernstian interfacial potential, affect the channel conductivity of the TFT, in addition to the Nernstian interfacial potential. A platform based on the charge-coupling extended-gate a-InGaZnO-based TFT is suitable for an ultrasensitive and flexible biosensing system to monitor not only pH but also biomarkers such as glucose in sweat on skin in the future.

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Ions and ionic biomolecules are powerful candidates for transmitting electrochemical information in the body, making them a useful label-free biosensing technology. A solution-gate field-effect transistor (FET) is based on the potentiometric measurement of ions at the gate surface of the FET and enables the direct detection of intrinsic ionic or biomolecular charges without labeling materials. In the last few decades, an ion-sensitive FET (ISFET) has been commonly used as the basic structure of ion sensors1,2 and potentiometric biosensors,3–9 with various semiconductor materials used as the channel, including both inorganic and organic materials.10–15 The gate insulator of an ISFET is composed of oxide thin films of, for example, SiO2 and Ta2O5, the surface of which has hydroxyl groups in a medium, provided the original work is properly cited. [DOI: 10.1149/2.0321813jes]

Experimental

Design of ISTFT.—An extended-gate ISTFT was prepared to design the charge-coupling circuit, the sensing electrode of which was separated from the a-InGaZnO-based ISTFT, as shown in Fig. 1. A change in ionic charges on the extended electrode with a fixed area is induced by pH variation and biological reactions, and the size of the gate channel of the TFT can be varied. The diameter (d) of the extended electrode with Ta2O5 (Fig. 1) was fixed to 18 mm (Sn = πd2/4), while the width (W) and length (L) of the TFT channel was varied to W × L (Sdc) = 50 × 50, 50 × 20, 50 × 10, 10 × 10,
and 10 × 5 (Fig. 1). All the thin films as the sensing electrode in this study were fabricated by sputtering (ULVAC, Inc.), and their thicknesses were 100 nm for Ta2O5, 80 nm for Au, and 20 nm for Cr. Ta2O5 was sputtered with 10% O2. A Cr film was used as the adhesive layer between Au and the oxide in all the experiments. Each electrode surface was cleaned by exposure to UV/ozone for 2 min. On the other hand, the electrical properties of an insulated-gate TFT with $W \times L = 30 \times 10$ for pH variation (Fig. S1 in Supplemental Material) were compared with those of the extended-gate TFT.

In the development of TFT devices (Fig. 1 or Fig. S1 in Supplemental Material), a-InGaZnO was sputtered with 5% O2, Mo or Au was used as conductive electrodes such as the source and drain, and SiO2 or Ta2O5 was thinly coated on the gate. The sputtered films were heat treated at 350°C for 1 h. The thicknesses of a-InGaZnO, the metal electrodes of source/drain, and the oxide gate insulator were 80 nm, 100 nm, and 100 nm, respectively.

**Measurement of electrical signal using ISTFT.**—The detection principle of the ISTFT sensor is based on the potentiometric detection of changes in the charge density at the gate of the TFT,13 on which specific bindings are induced for ionic or molecular recognition. Ionic charges at the sensing surface of the gate interact electrostatically with electrons in the channel across the thin layer of gate insulator charges at the sensing surface of the gate. Hence, specific bindings are induced for ionic or molecular recognition. Ionic charges at the sensing surface of the gate interact electrostatically with electrons in the channel across the thin layer of gate insulator. Positive charges based on H+ are induced on the insulated-gate TFT surface (SiO2, Ta2O5, etc.) induced a larger signal than that obtained from the previous data based on capacitive-coupling effect.20–22 The output signal when using the extended-gate ISTFT with $W \times L = 10 \times 10$, as shown in Fig. 2. The negative shift of $V_T$ with decreasing pH indicates an increase of H+ on the basis of the equilibrium reaction between H+ and OH− at the oxide film. Moreover, as shown in Fig. 3, the pH sensitivity was enhanced to approximately 313 mV/pH in the case of $W \times L = 10 \times 5$ and was then evaluated by changing the reciprocal of the channel area ($1/S_{\text{WL}}$, where $S_{\text{WL}} = W \times L$) from 4 × 10−4 to 2 × 10−2 [$\text{cm}^2$], although the pH sensitivities were near the Nernstian response for the extended-gate TFTs with $W \times L = 50 \times 50$ and 50 × 20. This resulted from the charge-coupling circuit and produced a larger signal than that obtained from the previous data based on the capacitive-coupling effect.20–22 The output signal when using the extended-gate TFT was several times larger than that when using the insulated-gate TFT (Fig. S3 in Supplemental Material). The enhancement of the pH sensitivity based on the charge-coupling circuit should also contribute to the improvement of detection sensitivity for biomolecules such as glucose.

Similarly to in the operation of a FET device, $I_D$ for an insulated-gate TFT in the unsaturated region can be generally described by

$$I_D = \mu C_{ox,TFT} \frac{W}{L} \left[ (V_G - V_T) V_D - \frac{1}{2} V_D^2 \right],$$  \[3\]

where $I_D$ is the drain current, $\mu$ is the electron mobility in the channel, $C_{ox,TFT}$ is the gate oxide capacitance, $W$ is the thickness, $L$ is the channel length, $V_D$ and $V_G$ are the drain-source and gate-source voltages, respectively, and $V_T$ is the threshold voltage, which can be $\Delta V_T = \frac{Cox}{\mu L^2} \cdot \frac{\partial}{\partial pH} \Delta pH$.

**Results and Discussion**

As shown in our previous paper,13 the $V_G$–$I_D$ electrical characteristics were obtained in solutions with various pH values using phosphate buffer solutions (PBSs) were prepared with pH values from 6.41 to 7.43 by controlling the mixing ratio of Na2HPO4 to KH2PO4 (Wako Pure Chemical Industries, Ltd.) with a fixed area ($d = 18$ mm) was induced by pH variation, and the channel size ($W \times L$) of the TFT was varied. As a result, $\Delta V_T$ was found to be approximately 198 mV/pH, depending on the pH, from the $V_G$–$I_D$ electrical characteristic obtained using the extended-gate ISTFT with $W \times L = 10 \times 10$, as shown in Fig. 2.

**Figure 1.** Structure of charge-coupling extended-gate InGaZnO-based TFT. The Ta2O5 film was used for pH sensing.
Figure 2. $V_G-I_D$ electrical characteristics of charge-coupling extended-gate InGaZnO-based TFT in various solutions. The pH of the measurement solution was varied from 4.01 to 9.18. The shift in $V_G$ was evaluated as $\Delta V_G$ while varying the pH at a constant $I_D$ of 0.6 μA.

expressed by\textsuperscript{23}

$$V_T = E_{\text{ref}} - \psi_0 + \chi^\text{tot} - \frac{\phi_{\text{d}}}{q} - \frac{Q_{\text{it},\text{Ex}} + Q_{\text{f},\text{Ex}} + Q_{\text{b},\text{Ex}}}{C_{\text{ox}}},$$

where $E_{\text{ref}}$ is the reference electrode potential relative to a vacuum, $(\psi_0 + \chi^\text{tot})$ is the interfacial potential at the electrolyte/gate oxide interface (the factor $\chi^\text{tot}$ is the surface dipole moment of the solution, which can be considered to be constant), $\phi_{\text{d}}$ is the a-InGaZnO electron work function, $Q_{\text{it},\text{Ex}}, Q_{\text{f},\text{Ex}},$ and $Q_{\text{b},\text{Ex}}$ are the charge of the interface traps, the fixed oxide charge, and the bulk depletion charge, respectively, and $\phi_f$ is the Fermi potential of a-InGaZnO. However, Equation 4 gives $V_T$ for an insulated-gate TFT; therefore, this equation should be modified as follows for an extended-gate TFT:

$$V_T = E_{\text{ref}} - \psi_0 + \chi^\text{tot} - \frac{\phi_{\text{d}}}{q} - \frac{Q_{\text{it},\text{TFT}} + Q_{\text{f},\text{TFT}} + Q_{\text{b},\text{TFT}}}{C_{\text{ox}}} + 2\phi_f,$$

where $C_{\text{ox}}$ is the capacitance of the extended Ta$_2$O$_5$ film, and $Q_{\text{it},\text{TFT}}, Q_{\text{f},\text{TFT}},$ and $Q_{\text{b},\text{TFT}}$ are the charge of the interface traps between the extended Ta$_2$O$_5$ and C/Au films and the fixed oxide charge of the extended Ta$_2$O$_5$ film, respectively. Moreover, we consider the effect of ionic charges at the extended oxide surface on the channel of the TFT device. As shown in Fig. 4, the Nernstian interfacial potential depends on $C_{\text{ox}}$ (Equations 1 and 2), which can be directly detected as the output voltage using the insulated-gate TFT, while part of the ionic charges (dotted circle) at the solution/electrode interface, which are unrelated to the Nernstian interfacial potential, are hypothesized to affect the channel of the TFT. That is, the change in such ionic charges of $\Delta Q_{\text{surf,TFT}}$, which does not directly contribute to the formation of the electrical double layer, are assumed to induce the change in such ionic charges at the extended electrode contribute to the change in $V_T$ of a TFT device through the charge-coupling circuit, although it was difficult to determine the mechanism how $\Delta Q_{\text{surf,TFT}}$ was induced. Therefore, Equation 5 is modified as

$$V_T = E_{\text{ref}} - \psi_0 + \chi^\text{tot} - \frac{\phi_{\text{d}}}{q} - \frac{Q_{\text{it},\text{TFT}} + Q_{\text{f},\text{TFT}} + Q_{\text{b},\text{TFT}}}{C_{\text{ox}}} + 2\phi_f,$$

where the electrical double layer is formed by the layer of charges at the oxide surface and the layer of charges based on counter ions. Thus, the rate of change affects the channel of the TFT ($\Delta V_T$) is expressed by Equation 6 using the total charge $Q_{\text{surf,total}}$, which is composed of the charge of the electrical double layer $Q_{\text{surf,DL}}$ and $Q_{\text{surf,TFT}}$. From Equations 6 and 7, $\Delta V_T$ caused by pH variation is calculated as

$$\Delta V_T = \Delta \psi_0 - \frac{\delta Q_{\text{surf}}}{C_{\text{ox}}} + \frac{Q_{\text{surf,DL}}}{C_{\text{ox}}},$$

In particular, $C_{\text{ox},\text{TFT}} \ll C_{\text{ox},\text{Ex}}$ in this study; therefore, Equation 8 is modified as

$$|\Delta V_T| = \frac{\delta Q_{\text{surf}}}{C_{\text{ox},\text{TFT}}},$$

where $\Delta \psi_0$ is the change in the interfacial potential corresponding to the Nernstian response shown in Equation 1, $\varepsilon_{\text{ox}}$ is the permittivity of the oxide membrane (the relative dielectric constant of SiO$_2$ = 3.9),\textsuperscript{34} and $s_{\text{ox}}$ is the thickness of the oxide membrane. If $C_{\text{ox},\text{TFT}} \gg C_{\text{ox},\text{Ex}}$, sufficient value of $\Delta Q_{\text{surf,total}}$ is not expected because the surface area of the extended electrode is assumed to be too small in most cases. Considering Equations 10 and 10, the output voltage $|\Delta V_T|$ increases with decreasing the channel size ($W \times L$), in addition
to the Nernst response (ΔΨN). The expected trend in the output voltage was in good agreement with the higher pH responsivity with the decrease in W × L shown in Fig. 3. Thus, the effect of Ψsurf,TFT on the amplification of the output signal cannot be neglected in the charge-coupling effect using the extended-gate TFT sensor. This is because ΔQsurf,total was sufficiently large to induce ΔQsurf,TFT affecting the channel conductivity of the TFT. That is, the surface area of the extended electrode was much larger than that of the channel, and then ΔQsurf,TFT should affect the channel of the TFT. In fact, the pH sensitivities were near the Nernstian response for the extended-gate TFTs with Ssw/SWL = 2.5 ∼ 6.4 × 10⁴ (W × L = 50 × 50 and 50 × 20), but they were enhanced for the extended-gate TFTs with Ssw/SWL more than 1.2 × 10⁷ (W × L = 50 × 10, 10 × 10, and 10 × 5), as shown in Fig. 3. Also, the insulated-gate a-InGaZnO-based TFT devices merely showed pH sensitivity (62.9 mV/pH) near the Nernstian response (Fig. S3 in Supplemental Material). This is assumed to be because the effect of ΔQsurf,TFT on the channel of the TFT was rather small for the insulated-gate TFT compared with that for the extended-gate TFT because the surface area of the sensing electrode was mostly consistent with W × L for the channel in the case of the insulated-gate TFT. That is, the size of sensing electrode can be controlled by using the extended-gate structure. Thus, the design of a device structure to yield a large value of ΔQsurf,TFT should contribute to increasing the detection sensitivity of the ISTFT to pH variation, although some uncertainties of parasitic capacitances, such as those between channel oxide and drain, source, semiconductor, and substrate, can’t be neglected for a qualitative analysis of electrical signal as the effect of nonidealities caused by device geometries.39

Moreover, ΔΨN was calculated using the pH sensitivity shown in Fig. 3 and Equations 9 and 10. From the data in Fig. 3,

\[
\text{pH sensitivity} = |\Delta V_T| = 51.7 + \frac{1.3 \times 10^4}{W \times L} \text{[mV/pH],}
\]

while from Equations 9 and 10,

\[
\text{pH sensitivity} = |\Delta V_T| = \frac{\beta}{\beta + 1} \times 59.1
\]

Comparing Equations 11 and 12, ΔΨsurf,total was calculated to be 1.1 × 10⁻¹² [C/pH] and β was calculated to be 7.1, where \( N_S = 1.0 \times 10^{25} \), \( K_a = 10^{-2} \), and \( K_b = 10^{-4} \) for Ta₂O₅ of the extended-gate film were used in Equation 2, resulting in \( C_{DL} = 1.8 \times 10^{-6} \) [F]. Therefore, ΔΨN was calculated to be 1.2 × 10⁻⁵ using Equation 7, where \( Q_{surf,DL} = C_{DL} \times \Delta \phi_N = 1.8 \times 10^{-6} \times 0.0517 \). Thus, ΔΨN can be considered as a coupling factor that indicates the improvement of the detection sensitivity of ions or biomolecules, although further experiments are needed to evaluate this factor, which has not been reported elsewhere.

**Conclusions**

In this study, we determined the effect of ionic charges at the sensing electrode, which were unrelated to the Nernstian interfacial potential, on the enhancement of detection sensitivities for glucose and pH. For this purpose, a charge-coupling extended-gate TFT structure, which is based on the concept of capacitive-coupling method, was employed to control the ratio between the size of the extended electrode and the channel size of the TFT. It was found that the pH sensitivity was amplified with decreasing channel size (capacitance) of the TFT gate, while maintaining a constant surface area of the extended electrode with an oxide membrane. This result indicates that the output signal can be controlled via the ratio of the product of the coupling factor and the change in the total number of charges at the extended electrode to the channel capacitance (Csurf,TFT), in addition to the change in the Nernstian interfacial potential at the electrical double layer. In fact, the pH sensitivity was enhanced to 313 mV/pH, greatly exceeding the Nernstian response (59.1 mV/pH at 25°C). A platform based on the charge-coupling-gate a-InGaZnO-based TFT is suitable for an ultrasensitive and flexible biosensing system to monitor not only pH but also biomarkers such as glucose in sweat on skin in the future.

Such potentiometric sensors can also be applied as a bioanalytical tool in the field of medicine and in vitro diagnostics (IVD). Glucose sensors are widely used to monitor glucose levels in blood for diabetic
mellitus patients. Most glucose-sensing devices utilize the enzymatic reaction between glucose oxidase (GOD) and glucose owing to the high selectivity, convenience, and low cost, the detection principle of which can be used to analyze the glucose level with a sensitivity of about 100 mg/dl (about 5.5 mM) in blood. However, a conventional glucose sensor has several disadvantages; commercially available glucose sensors require blood sampling to measure blood glucose, which may be painful, and their detection sensitivity is insufficient at low glucose levels in biological fluids such as sweat, which can be used for noninvasive glucose measurement instead of blood samples. In fact, the glucose levels in sweat are about one-hundredth to one-tenth of those in blood. Therefore, the charge-coupling effect can be used for noninvasive glucose measurement instead of blood samples. In fact, the glucose levels in sweat are about one-hundredth to one-tenth of those in blood. Therefore, the charge-coupling effect leads to a highly sensitive biosensor, which enables the monitoring of low levels of glucose in biological fluids such as sweat. Additionally, TFT-based flexible devices can be used as a noninvasive biosensor by placing them on the skin similarly to a plaster in the future.

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References

1. P. Bergveld, IEEE Trans. Biomed. Eng., BME-19, 342 (1972).
2. M. Esashi and T. Matsuo, IEEE Trans. Biomed. Eng., BME-25, 184 (1978).
3. P. Fromherz, A. Offenhausser, T. Vetter, and J. Weis, Science, 252, 1290 (1991).
4. T. Sakata and Y. Miyahara, Angew. Chem., Int. Ed., 45, 2225 (2006).
5. T. Sakata and Y. Miyahara, Anal. Chem., 80, 1493 (2008).
6. T. Sakata, M. Ihara, I. Makino, Y. Miyahara, and H. Ueda, Anal. Chem., 81, 7532 (2009).
7. J. M. Rothberg, W. Hinz, T. M. Rearick, J. Schultz, W. Mileski, M. Davey, J. H. Leaman, K. Johnson, M. Milgrew, M. Edwards, J. Hoon, J. F. Simons, D. Marran, J. W. Myers, J. F. Davidson, A. Branting, J. R. Nobile, B. P. Pac, D. Light, T. A. Clark, M. Huber, J. T. Branciforte, I. B. Stoner, S. E. Cawley, M. Lyons, Y. T. Pa, N. Homer, M. Sedova, X. Miao, B. Reed, J. Sabina, E. Feierstein, M. Schorn, M. Alanjary, E. Dimalanta, D. Dressman, R. Kasinskas, T. Sokolsky, J. A. Fidanza, E. Namsaraev, K. J. McKernan, A. Williams, G. T. Roth, and J. Bustillo, Nature, 475, 348 (2011).
8. T. Sakata, A. Saito, J. Mizuno, H. Sagimoto, K. Noguchi, E. Kikuchi, and H. Inui, Anal. Chem., 85, 6633 (2013).
9. H. Yang, M. Honda, A. Saito, T. Kajisa, Y. Yanase, and T. Sakata, Anal. Chem., 89, 12918 (2017).
10. D. Sarkar, W. Liu, X. Xie, A. C. Anselmo, S. Mitragotri, and K. Banerjee, ACS Nano, 8, 3992 (2014).
11. T. Ikaz, T. Sakata, Y. Miyazawa, T. Kajisa, A. Kromka, and B. Rezek, Diamond and Related Materials, 60, 87 (2015).
12. C. Reimer-Rozmann, M. Larioka, C. Nowak, and W. Knoll, Biosens. Bioelectron., 70, 21 (2015).
13. T. Sakata, K. Nishimura, Y. Miyazawa, A. Saito, H. Abe, and T. Kajisa, Anal. Chem., 89, 3901 (2017).
14. T. Minamiki, Y. Sasaki, S. Tokito, and T. Minami, ChemistryOpen, 6, 472 (2017).
15. N. T. Tung, P. T. Tue, T. T. N. Lien, Y. Ohno, K. Maehashi, K. Matsumoto, K. Nishigaki, M. Biyani, and Y. Takamura, Sci. Rep., 7, 17881 (2017).
16. L. Bousse, M. Mostashref, B. Van Der Shoot, N. F. de Rooij, P. Gimmel, and G. Göpel, J. Coll. Interface Sci., 147, 229 (1991).
17. T. Sakata and A. Saito, ECS J. Solid State Sci. Technol., 7, Q3094 (2018).
18. T. Sekitani, U. Zschieschang, H. Klauk, and T. Someya, Nature Mater., 9, 1015 (2010).
19. M. Kaltenbrunner, T. Sekitani, J. Reeder, T. Yokota, K. Kuribara, T. Tokuhara, M. Drack, R. Schwindauer, I. Graz, S. Bauer-Gogonea, S. Bauer, and T. Someya, Nature, 499, 458 (2013).
20. K. B. Partri, A. J. Yeh, A. S. Y. Poon, and H. S. P. Wong, IEDM12, 557 (2012).
21. C. Duarte-Guevara, F.-L. Lai, C.-W. Cheng, B. Reddy Jr., E. Salim, V. Swaminathan, Y.-K. Tsui, H. C. Tuan, A. Kalinitsky, Y.-S. Liu, and R. Bashir, Anal. Chem., 86, 8359 (2014).
22. K. Takechi, S. Iwamatsu, T. Yahagi, Y. Abe, S. Kobayashi, and H. Tanabe, Jpn. J. Appl. Phys., 53, 076702 (2014).
23. P. Bergveld, Sens. Actuators, 8, 109 (1985).
24. L. Bousse, J. Chem. Phys., 76, 5128 (1982).
25. D. Hrame, L. Bousse, J. Shott, and J. Meindl, IEEE Trans. Electron Devices, 34, 1790 (1987).
26. M. Yu and T. M. Volkovich, Russian Journal of Electrochemistry, 38, 935 (2002).
27. T. Sakata, M. Kamahori, and Y. Miyahara, Jpn. J. Appl. Phys., 44, 2954 (2005).
28. Y. Maekawa, Y. Shibuta, and T. Sakata, Jpn. J. Appl. Phys., 52, 127001 (2013).
29. Y. Maekawa, Y. Shibuta, and T. Sakata, Chem. Phys. Lett., 619, 152 (2015).
30. Y. Maekawa, Y. Shibuta, and T. Sakata, ChemElectroChem, 1, 1516 (2014).
31. B. M. Lowe, Y. Maekawa, Y. Shibuta, T. Sakata, C.-K. Skylaris, and N. G. Green, Phys. Chem. Chem. Phys., 19, 2687 (2017).
32. B. M. Lowe, C.-K. Skylaris, N. G. Green, Y. Shibuta, and T. Sakata, Jpn. J. Appl. Phys., 57, 04FM02 (2018).
33. B. M. Lowe, C.-K. Skylaris, N. G. Green, Y. Shibuta, and T. Sakata, NanoJoule, 10, 8650 (2018).
34. J. Robertson, Reports Prog. Phys., 69, 327 (2006).
35. Y. Liu, P. Georgiou, T. Prodromakis, T. G. Constandinou, and C. Tourmazou, IEEE Trans. Electron Devices, 58, 4414 (2011).
36. S. J. Updike and G. P. Hicks, Nature, 214, 986 (1967).
37. C. Steckling, Chem. Soc. Rev., 6, 215 (1977).
38. J. D. Newman and A. P. P. Turner, Biosens. Bioelectron., 20, 2435 (2005).
39. J. Moyer, D. Wilson, I. Finkelstein, B. Wong, and R. Potis, Diabetes Technol. Ther., 14, 398 (2012).