Glocuse detection with a commercial MOSFET using ZnO nanowires extended gate

Syed M. Usman Ali, Omer Nour, Magnus Willander and Bengt Danielsson

Syed M. Usman Ali, Omer Nour, Magnus Willander and Bengt Danielsson, Glocuse detection with a commercial MOSFET using ZnO nanowires extended gate, 2009, IEEE Transaction on Nanotechnology, (8), 6, 678-683.
http://dx.doi.org/10.1109/TNANO.2009.2019958

Glucose Detection With a Commercial MOSFET Using a ZnO Nanowires Extended Gate

Syed Muhammad Usman Ali, Omer Nur, Magnus Willander, and Bengt Danielsson

***Abstract*—ZnO nanowires were grown on Ag wire with a diameter of $\sim$250 μm and used in an electrochemical sensor. The enzyme glucose oxidase (GOD) was immobilized on the ZnO nanowires, and the Ag wire was connected directly to the gate of a MOSFET. Upon exposure to glucose (1–100 μM), the electrochemical response from the GOD induced a stable measurable voltage change on the gate leading to a strong modulation of the current through the MOSFET. For a sensor with uniform ZnO nanowires functionalized with GOD, a fast response time of less than 100 ms was demonstrated. The effect of the uniformity of the ZnO nanowires on the sensing property was also investigated. The extended-gate arrangement facilitated glucose detection in small sample volumes, and made it possible to demonstrate the present sensor concept using a standard low-threshold MOSFET. The extended-gate MOSFET sensor approach demonstrates the possibility and potential of the use of nanostructures coupled to standard electronic components for biosensing applications.**

***Index Terms*—Biomedical transducers, biosensor, electrochemical devices, MOSFETs.**

I. Introduction

DETECTION of biological or biochemical processes is of utmost importance for medical and biotechnological applications. However, converting the biological signal to an easily processed electronic signal is challenging due to the complexity of connecting an electronic device directly to a biological environment. Electrochemical biosensors provide an attractive means to analyze the content of a biological sample due to the direct conversion of a biological event to an electric signal. Over the past decades, several sensing concepts and related devices have been developed.

The area of biosensors started to be active with the introduction of the first generation of glucose oxidase (GOD) biosensors in 1962 [1]. This GOD sensor concept is still the most widely used, although many improvements (generations) have been added since 1960's [2]. As exemplified by the glucose sensor, electrochemical biosensors do not suffer the drawback of high sensor setup complexity and cost. This is due to their close link to developments in low-cost production of microelectronic circuits and their easy interface with normal electronic readout and processing. Other inherent advantages of electrochemical biosensors are their robustness, easy miniaturization, excellent detection limits, also with small analyte volumes, and ability to be used in turbid biofluids with optically absorbing and fluorescing compounds [3], [4]. However, several aspects such as low operational stability and interference problems in complex sample matrices could be considered to have held back the emergence of additional breakthrough applications build on electrochemical biosensing.

Many potentiometric devices are based on various forms of FET devices to measure pH changes, selective ion concentrations, and the kinetics of biocatalytic reactions involving enzymes [5]. The conversion of a FET into a sensing device normally involves the replacement of the metal gate electrode by a biochemically sensitive surface (e.g., an analyte-selective membrane, an enzyme layer or an ion-conductive solution, etc.), which is brought into contact with the analyte solution [6]. Also present in the analyte solution is a reference electrode, which completes the circuit via the gate voltage bias [7], [8]. One of the most popular methods for the construction of FET-based biosensing devices is the immobilization of enzymes at the gate surface of pH-sensitive ion-sensitive FET (ISFET) devices. In general, nanowires belong to a growing family of nanostructures, which also includes nanotubes, nanoparticles, nanowires, nanobelts, nanosprings, and more [9]–[14]. Nanowires are attractive for their versatile roles in bioelectronics and nanoelectronics applications, and they are increasingly being used as building blocks for biosensing purposes. The use of nanomaterials has allowed the introduction of many new signal transduction technologies in biosensors resulting in improved sensitivity and performance. Because of their submicrometer dimensions, nanosensors, nanoprobes, and other nanosystems have allowed simple and rapid analyses *in vivo*. Their implementation as highly sensitive electrodes is one obvious example, such as the platinum electrode network proposed by Wang *et al.* for glucose detection [15]. Among the nanostructures, ZnO is of special interest to biological sensing due to many favorable properties. Being next to 1-D nanostructure, ZnO nanorods have many unique advantages including high surface-to-volume ratio, good chemical stability, and easy fabrication. They are nontoxic, electrochemically active, and have a high electron communication capability. Recently, ZnO-based nanomaterials have been used as biocompatible materials in gas and pH sensors [16], [17]. When considering glucose detection, it is important to note that

Manuscript received December 22, 2008; revised March 6, 2009. First published April 3, 2009; current version published November 11, 2009. The review of this paper was arranged by Associate Editor B. Nelson.

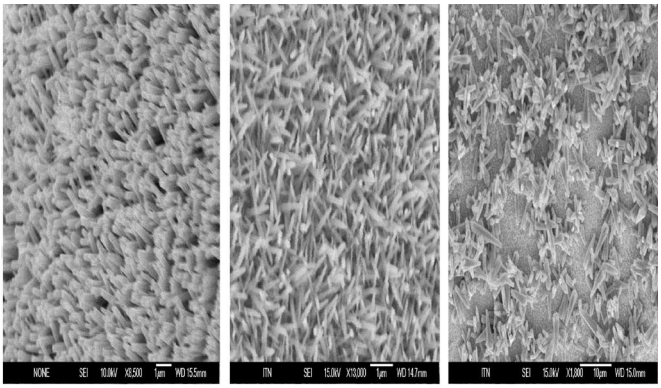
S. M. Usman Ali and O. Nur are with the Department of Science and Technology, Linkoping University, SE-60174 Norrköping, Sweden (e-mail: syeal@itn.liu.se; omeno@itn.liu.se).

M. Willander is with the Department of Science and Technology, Linköping University, SE-601 74 Norrköping, Sweden, and also with the Department of Physics, University of Gothenburg, SE-41296 Gothenburg, Sweden (e-mail: magwi@itn.liu.se).

B. Danielsson is with the Division of Pure and Applied Biochemistry, Lund University, SE-22100 Lund, Sweden (e-mail: bengt.danielsson@tbiokem.lth.se).

Color versions of one or more of the figures in this paper are available online at http://ieeexplore.ieee.org.

Digital Object Identifier 10.1109/TNANO.2009.2019958



(a) (b) (c)

Fig. 1. (a) Typical SEM image of ZnO nanowires grown on 250-μm Ag wire using low temperature chemical growth. The figure shows that the diameter of the nanowires is in the range of 250–300 nm. (b) Case of nonaligned uniformly distributed nanowires. (c) Case of nonaligned, nonuniformly distributed nanowires.

there is a large difference in the isoelectric points (IPs) of ZnO and glucose oxidase, which implies that glucose oxidase can easily be electrostatically immobilized on ZnO nanostructures. Hence, ultrasensitive glucose sensors can be demonstrated using ZnO nanowires [18]. The electrostatic interaction between ZnO and GOD is especially favorable around neutral pH [19], [20].

In this paper, we report how instead of growing ZnO nanowires on the gate area inside the transistor (e.g., on the AlGaN/GaN HEMT device), ZnO nanowires can be integrated on the surface of a Ag wire (with a diameter of around 250 μm) as an extended gate [21]. In this way, the chemically sensitive gate is then separated from the rest of the transistor construction, and the sensing area increases significantly as compared to gate areas of some published sensors based on transistors, e.g., HEMT [22]. Thereby, the biosensor construction is much facilitated as the enzyme can be readily immobilized on the wire, and applied in a variety of different probes or flow systems designs without problems arising from, e.g., encapsulation of the electronics, etc. In addition, we report on the effect of the uniformity and vertical orientation of ZnO nanowires on response time of the sensor.

II. Experimental

A. ZnO Nanowires Growth

A 3-cm-long clean, straight piece of Ag wire (250 μm in diameter) was first rinsed with acetone followed by rinsing in deionized water, and then, it was dried at room temperature. To grow ZnO nanowires on the Ag wire, low-temperature chemical approach was adopted [23]. First, the sliver wire was dipped into a seed solution for 2 min, and then dried in air. This procedure was repeated twice. The seed solution contained 0.025 M zinc nitrate and 0.025 M hexamethylenetetramine [HMT, ($C_6H_{12}N_4$)]. The solution was kept at 90 °C during ZnO nanowires growth. Subsequently, the wire was washed by distilled water and dried at room temperature. Typical ZnO nanowires grown on the Ag wires using this procedure are shown in Fig. 1(a). As clearly seen, ZnO nanowires of 250–300 nm diameters with uniform density and spatial distribution was grown. These nanowires were relatively perpendicular to the surface of the Ag wire. The morphological and structural characteristics of the grown nanowires can be controlled by adjusting the growth process parameters such as the concentration of the seed solution, the reagent stoichiometry, the temperature, and the pH of growth solution [24], [25]. The results presented here were achieved by varying the temperature and the time during the growth process, while the pH and concentration of reactants were kept constant. Some samples were grown with parameters differing from the parameters that produced the nanowires shown in Fig. 1(a). The other ZnO samples are shown in Fig. 1(b) and (c). For the second sample shown in Fig. 1(b), the solution was kept at 92 °C in oven for 90 min, and the result was a nonaligned but uniformly distributed nanowires. Similarly, for the third sample shown in Fig. 1(c), the solution was kept at 95 °C in oven for two-and-a-half hour, and the result was nonaligned nonuniformly distributed ZnO nanowires. All these samples were tested for their glucose sensing response.

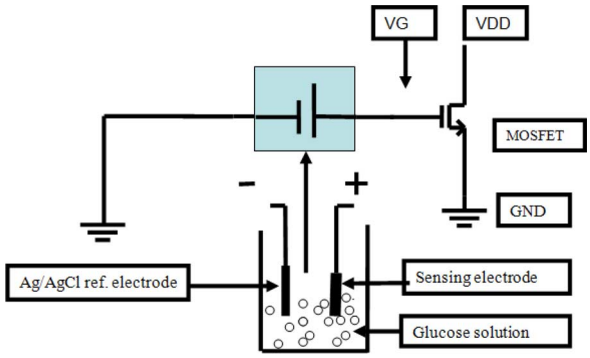


Fig. 2. Schematic diagram illustrating the configuration used for glucose detection with MOSFET using extended-gate functionalized ZnO nanowires as working electrode and Ag/AgCl as a reference electrode.

GOD solution, 10 mg/ml, was prepared in 10 mM phosphate buffered saline, pH 7.4, using GOD (E.C. 1.1.3.4) type X-S, 100 U/mg (Sigma Aldrich, St. Louis, MO). GOD was electrostatically immobilized by dipping the ZnO-nanowires-coated Ag wire into 5 μL of the enzyme solution of 1:1 ratio for 15 min at room temperature, and then, it was dried in air for more than 20 min. After completing these steps, the sensor [sample in Fig. 2(a)] was checked in 100 μL of 100 μM glucose solution potentiometrically with an Ag/AgCl as a reference electrode. A substantial response of $\sim$85 to 100 mV was observed. The output response depends on the GOD enzyme immobilization and the surface area dipped into the glucose solution. This implies that GOD immobilized on ZnO nanowires have reacted with the glucose and the electrons resulting from this reaction had been transferred to the ZnO.

The mechanism of electrochemical glucose sensors is based on an enzymatic reaction catalyzed by GOD according to the following equation:

$$\mathrm{H_2O} + \mathrm{O_2} + \mathrm{Glucose} \xrightarrow{\mathrm{GOD}} \beta\text{-gluconolactone} + \mathrm{H_2O_2}. \quad (1)$$

As the result of this reaction, β-gluconolactone and hydrogen peroxide are produced. These two products and the oxygen consumption can be used for the glucose determination. With H_2O availability in the reaction, gluconolactone is spontaneously converted to gluconic acid, which at neutral pH, form the charged products of gluconate $^-$ and proton$^+$, according to the following equation:

$$\beta\text{-Gluconolactone} \xrightarrow{\text{Spontaneous}} \text{Gluconate}^- + \text{H}^+. \quad (2)$$

This proteolytic product of the hydrolysis reaction of the β-gluconolactone to gluconic acid shown in (2), which results in a decrease of the medium pH that is usually used for the determination of the glucose concentration [26]. Another approach to assay glucose is based on electrochemical oxidation of hydrogen peroxide, which is generated according to (1). In this case, it is preferable to include redox mediators in the matrix of the immobilized enzymes to improve the electron transfer between the enzymes active centers and electrode surface [27]. ZnO nanowires possess an IP of 9.5 while the GOD has an IP equal to 4.2. This relatively large difference will lead to electrostatic attraction (immobilization) of the GOD on the ZnO nanowires surface. Moreover, as a traditional transparent conductor, single crystalline ZnO nanostructures have good conductivity, thus enhancing the direct electron transfer between the active sites of the enzyme (GOD in our case) and the sensor electrode (ZnO nanowires). These reasons contribute to the high sensitivity and high affinity of the biosensor constructed by ZnO nanowires. This response, i.e., glucose reacting with GOD immobilized on ZnO surface, was observed by many others using different approaches [19], [20], [28], [29].

B. Measurements and Results

After this experiment, we used a highly sensitive n-channel zero threshold ($V_{\text{th}} = 0$ V) ALD/110900 commercial n-MOSFET (Advanced Linear Devices, Sunnyvale, CA), which can operate in precision zero threshold mode. This transistor was integrated with the extended-gate sensor together with an Ag/AgCl electrode and connected to a Keithley 2602 unit, as schematically shown in Fig. 2. In addition, a pH meter (Model 744, Metrohm) was used to measure the potentiometric output voltage of the different ZnO nanowires sensors presented here. Moreover, time response measurements were also performed to study the stability. For the time response measurements, a model 363 A potentiostat/galvanostat (EG and G, Inc., Idaho Falls, ID) was used. The working electrode (ZnO nanowires) is negatively charged due to oxidation [see (1) and (2)]. The gate voltage must be positive in order to invert carriers at the n-MOSFET channel and observe the drain current modulations. The output voltage can be made positive by interfacing instrumentation amplifier in an inverting mode with unity gain between the sensor output and gate terminal of the MOSFET. If a p-channel MOSFET is used, then there is no need for an instrumentation amplifier interfacing.

Different device configurations were tested in order to distinguish the behavior and role of the ZnO nanowires in the sensing process. This test was performed by adopting a simple potentiometric measurement. Beside the GOD/ZnO on Ag wire configuration, two other different sensor configurations were tested for comparison, a bare Ag wire and GOD/Ag wire. Upon exposure of the bare Ag wire to 100 μM glucose, a voltage of around 65 mV was immediately observed. This signal, however, decayed continuously and reached to reversed polarity in less than 50 s, as shown in Fig. 3(a). It is worth mentioning that the

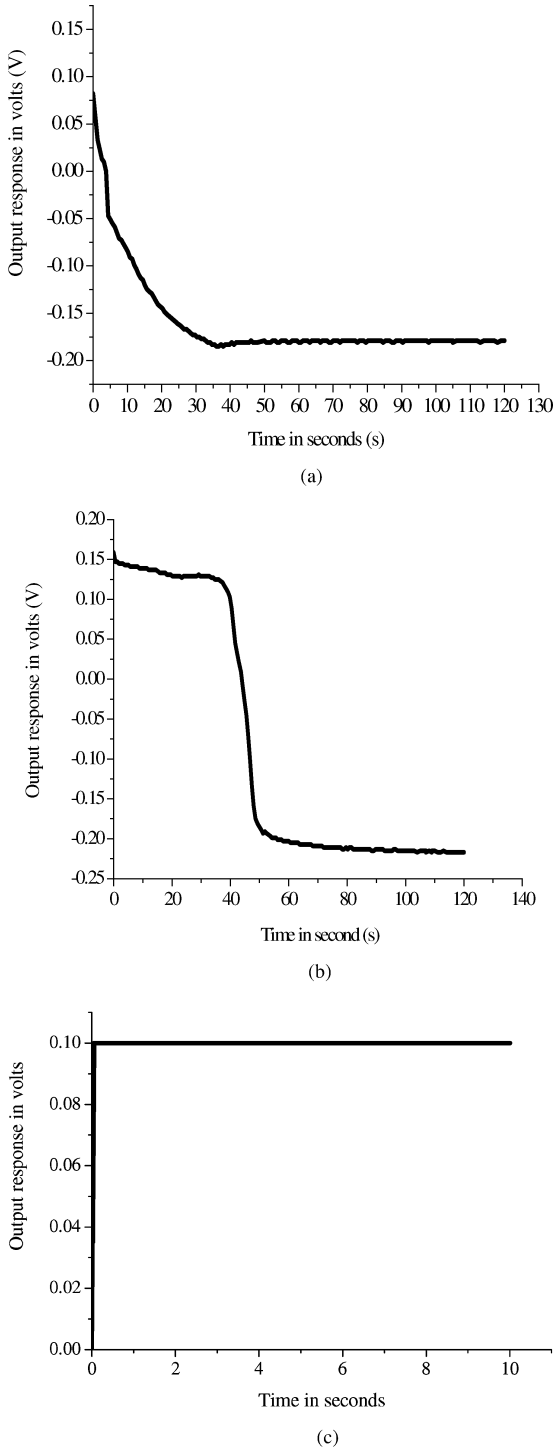


Fig 3. (a) Response of bare electrode without GOD in 100 μM glucose solution. (b) Response of bare electrode with GOD enzymes in 100 μM glucose solution. (c) Stable response of sensor electrode (Ag + ZnO + GOD) 100 μM glucose solution.

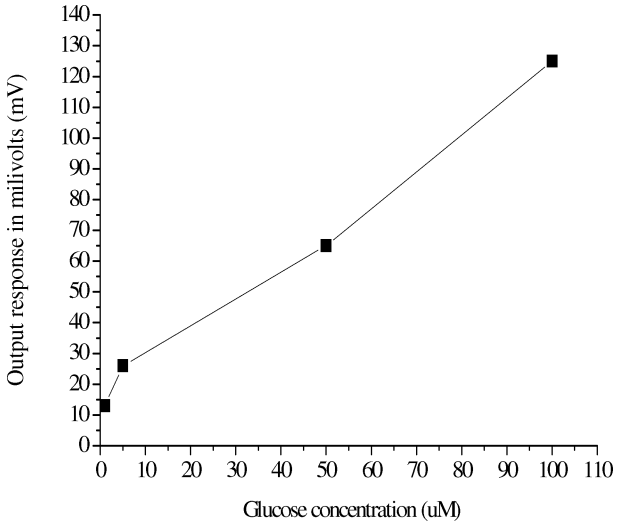


Fig. 4. Potentiometric response of GOD/ZnO sensor for different glucose concentrations (1–100 μM).

response decreased as the glucose concentration was decreased (for 1 μM glucose, an initial voltage of 25 mV was observed). This implies that a nonlinear relation between the response and concentration existed when bare Ag is used as the working electrode. When the GOD/Ag sensor configuration was tested, a different response was observed. Although it was also decaying, the sensor was stable for around 40 s, as shown in Fig. 3(b). Moreover, both these configurations showed no change of the pH during the sensing process. The same potentiometric test was performed for the GOD/ZnO/Ag working electrode, and the result showed stable behavior. The difference in the stability performance between the three different electrodes configurations is mainly due to the fact that in the case of the GOD/ZnO nanowires, the surface-to-volume ratio of the sensor in contact with the analyte is much larger than the other two cases; also, there is a large difference between the isoelectric points (IEPs) of ZnO and GOD, which provide strong electrostatic interaction between them. This implies that the sensitivity and stability is increased very much in the case of GOD/ZnO configuration compared to the other two cases. Moreover, due to the presence of the GOD enzyme, the reaction and transport of electrons is sustained with a high sensitivity. In addition, another reason for making the experiments with a bare Ag wire and a GOD-coated Ag wire connected to the gate of the MOSFET was to see what could happen if the GOD/ZnO layer of the sensor was damaged. From Fig. 3(a) and (b), it can be seen that there is only a transient effect with these wires because of no appreciable reaction on the bare Ag wire, and with little enzyme on the bare Ag wire, the rapid degradation of immobilized enzymes results in an unstable sensor signal. We did not see any such instability behavior in the experiments with the GOD/ZnO/Ag wires [shown in Fig. 3(c)]. Only these wires gave sustained signals because of sufficient catalytic activity and a much better transduction mechanism compared to the other two cases [Fig. 3(a) and (b)].

Then, the relation between the glucose concentration and the induced voltage was further investigated using simple potentiometric measurements. Fig. 4 shows the variation of the induced

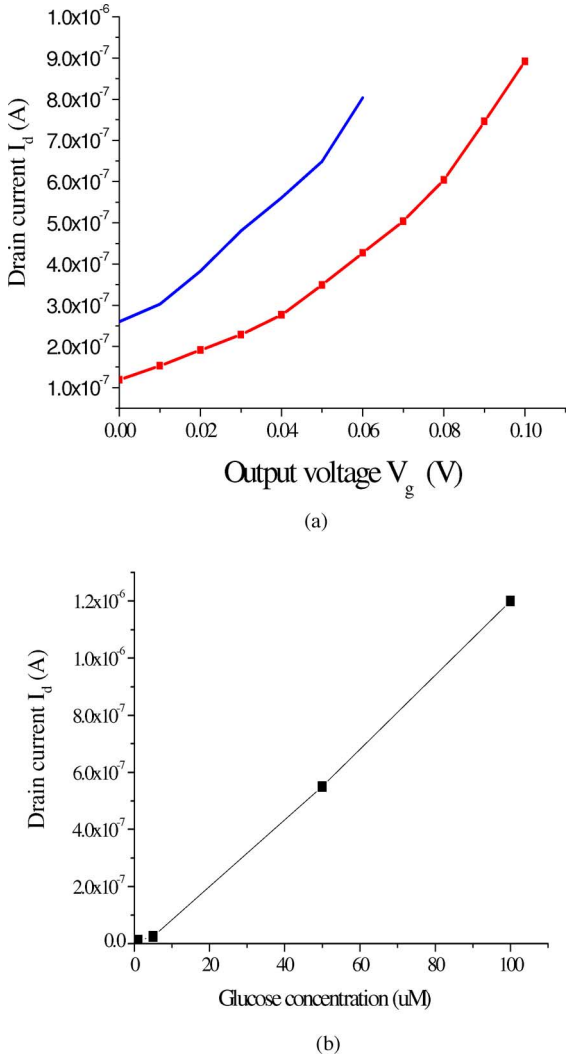


Fig. 5. (a) Typical drain current (I_D) versus gate voltage (V_G) for the extended-gate MOSFET, the upper curve (line) is for 50 μM glucose solution while the lower curve (dotted line) is for the case of 100 μM of glucose concentration. (b) Relation between the drain current and glucose concentration for a range of 1–100 μM glucose concentration.

voltage at the GOD/ZnO/Ag sensing electrode with the glucose concentration [30], [31]. This result shows a linear relationship between the induced voltage and the glucose concentration in the range investigated (1–100 μM). It is concluded that as the concentration of glucose increases, the output voltage of the sensor electrode increases.

The sensing electrode was interfaced to the gate of an n-channel MOSFET in order to further investigate the response to glucose detection. The MOSFET electrical characteristics for the GOD/ZnO/Ag extended-gate MOSFET sensor are shown in Fig. 5(a) and (b). The dc biasing was set as the standard transistor operation, i.e., $V_G = 0$ V and $V_{DD} = 2.0$ V.

The resulting offset leakage drain current was found to be around $I_D = 260$ nA, as shown in Fig. 5(a). This offset leakage current can be controlled by changing the drain supply voltage (V_{DD}). When the extended gate was immersed in 50 μM glucose solution, an induced voltage of about 60 mV was added to

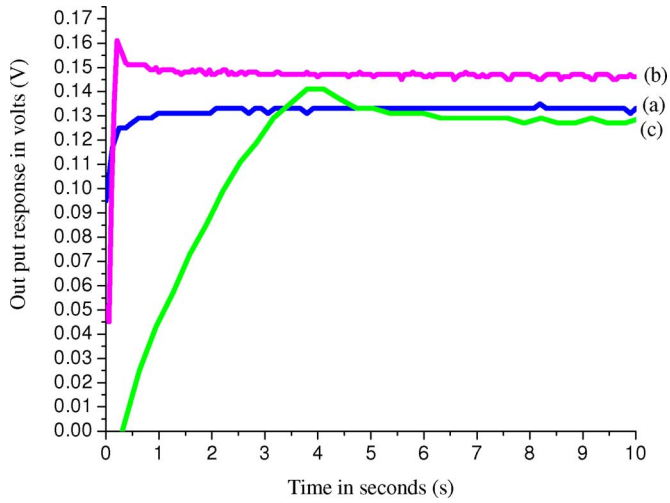


Fig. 6. Output responses of different types of nanowires grown by using different growth parameters in glucose solution (100 μM). The corresponding ZnO nanowires are shown in Fig. 1(a)–(c). (a) Response of vertically aligned uniformly distributed ZnO nanowires. (b) Response of nonaligned uniformly distributed nanowires. (c) Response of nonaligned, nonuniformly distributed nanowires.

the gate [see Fig. 5(a)]. As a result, a strong modulation of the drain current was observed. In this case, when the extended-gate transistor was stable, 0.540 μA was added to the drain current, as shown in Fig. 5(a). This increment is due to the reaction between the glucose and GOD/ZnO/Ag electrode leading to an electron transfer to the ZnO nanowires. It is important to mention that on contrary to the nonstable behavior of both the GOD/Ag and bare Ag wire, the modulation observed here prevail for a long time with no observable signal decay. The dependence of I_D on the glucose concentration is shown in Fig. 5(b). As evident form Fig. 5(b), the I_D increased as the glucose concentration is increased, as expected. It is also clear that the dependence of the I_D on the glucose concentration is showing a linear relationship. The reproducibility and long-term stability was evaluated by using five different sensor electrodes, all containing GOD/ZnO/Ag. After more than 20 successive measurements, the immobilized GOD/ZnO/Ag electrode has lost only 25% of its initial activity. Thus, the electrode retains 75% of the electrocatalytic activity of GOD, and prevents it from leaking out of the sensor.

The response time and stability of the extended-gate MOSFET electrochemical sensor was tested using samples with different ZnO nanowires distribution grown on the same type of Ag wire. We have chosen three different ZnO samples, the first one has uniform and rather vertically aligned ZnO nanowires, the second has uniform nonaligned ZnO nanowires, and the third one has nonuniform nonaligned ZnO nanowires, as shown in Fig. 1(a)–(c). For all three cases, the length of sensor wire dipped into the solution was the same, about 5 mm in length. Fig. 6 shows the response time versus the induced voltage for the three different sensors described earlier. As seen in the figure, for the uniform, well-aligned ZnO nanowires sensor (Fig. 6, curve a), a very stable signal and a response time less than 100 ms were observed. For the second case, when using uniform but not well-aligned ZnO nanowires, a rather fast response was also observed although the signal did not stabilize as fast as in the first case. In addition, an initial kink on the curve was observed (Fig. 6, curve b).

For the final case of using a nonaligned and nonuniform ZnO nanowires surface, the response was considerably slower compared to the two first cases. The response time to stabilize the response was more than 5.0 s, as shown in Fig. 6, curve c. It appears that the alignment and distribution of the ZnO nanowires play an role in the response time of the sensor configuration. This result is expected since the fluid contact to all parts of the ZnO functionalized surface depends on the ZnO nanowires distribution

III. Conclusion

In conclusion, we have demonstrated a robust glucose biosensor using functionalized ZnO nanowires coupled as an extended gate to a conventional low-threshold MOSFET. By immobilizing GOD on ZnO nanowires grown on Ag wire, a stable glucose-dependent modulation of the drain current of the MOSFET transistor was observed. The response time of well-aligned ZnO nanowires on the Ag wire is relatively fast; it was measured to be less than 100 ms. It has also been observed that if the nanowires are not well aligned and/or not uniformly distributed, then the response time is increased. This is observed because for the nonuniform nanostructure, more time to have contact with the fluids in question will be required. Hence, a slower response is expected as observed. These results demonstrate the capability of performing biologically relevant measurements with a functionalized gate externally connected at the terminal of commercial MOSFETs. Using this approach, we have measured small volumes of glucose solutions with concentrations ranging between 100 and 1 μM. The lowest detection limit, although not investigated here, is probably in nanomolar. The extended-gate MOSFET sensor concept presented here is robust, and opens the possibility of externally integrating nanosensing element to commercial transistors giving the advantages of simplicity and low cost. In addition, the extended gate makes it easier and more practical to sense elements when the available sample volume is relatively small. This simple method of fabricating ZnO/GOD biosensor can also be extended to immobilize other enzymes and other bioactive molecules on various nanostructures, and form versatile electrodes for biosensor studies.

References

[1] C. Lyons and L. C. Clark, Jr., "Electrode systems for continuous monitoring in cardiovascular surgery," *Ann. N. Y. Acad. Sci.*, vol. 102, pp. 29–45, 1962.

[2] A. P. Fang, H. T. Ng, and S. F. Y. Li, "A high-performance glucose biosensor based on monomolecular layer of glucose oxidase covalently immobilized on indium–tin oxide surface," *Biosens. Bioelectron.*, vol. 19, pp. 43–49, 2003.

[3] M. S. Wilson, "Electrochemical immuno-sensors for the simultaneous detection of two tumor markers," *Analytical Chem.*, vol. 77, pp. 1496–1502, 2005.

[4] P. D'Orazio, "Biosensors in clinical chemistry," *Clin. Chim. Acta*, vol. 334, pp. 41–69, 2003.

[5] F. Patolsky and C. M. Lieber, "Nanowires nanosensors," *Mater. Today*, vol. 8, pp. 20–28, 2005.

[6] A. Errachid, N. Zine, J. Samitier, and J. Bausells, "FET-based chemical sensor systems fabricated with standard technologies," *Electroanalysis*, vol. 16, pp. 1843–1851, 2004.
[7] B. Eggins, *Chemical Sensors and Biosensors in Analytical Techniques in the Sciences*. West Sussex: Wiley, 2002.
[8] M. J. Schoning and A. Poghossian, "Recent advances in biologically sensitive field-effect transistors (BioFETs)," *Analyst*, vol. 127, pp. 1137–1151, 2002.
[9] A. Merkoci, "Electrochemical biosensing with nanoparticles," *FEBS J.*, vol. 274, pp. 310–316, 2007.
[10] S. J. Park, T. A. Taton, and C. A. Mirkin, "Array-based electrical detection of DNA with nanoparticles probes," *Science*, vol. 295, pp. 1503–1506, 2002.
[11] A. K. Wanekaya, W. Chen, N. V. Myung, and A. Mulchandani, "Nanowire based electrochemical biosensors," *Electroanalysis*, vol. 18, pp. 533–550, 2006.
[12] C. M. Lieber and Z. L. Wang, "Functional nanowires," *MRS Bull.*, vol. 32, pp. 99–104, 2007.
[13] N. C. Tansli and Z. Q. Gao, "Nanoparticles in biomolecular detection," *Nano Today*, vol. 1, pp. 18–33, 2006.
[14] Y. Xiao, F. Patolsky, E. Katz, J. F. Hainfeld, and I. Willner, "Plugging into enzymes: Nanowiring of redox enzymes by a gold nanoparticles," *Science*, vol. 299, pp. 1877–1881, 2003.
[15] D. H. Wang, R. Kou, M. P. Gil, H. P. Jakobson, J. Tang, D. H. Yu, and Y. F. Lu, "Templated synthesis, characterization, and sensing application of macroscopic platinum nanowire network electrodes," *J. Nanosci. Nanotechnol.*, vol. 11, pp. 1904–1909, 2005.
[16] S. M. Al-Hilli, M. Willander, A. Öst, and P. Stålfors, "ZnO nanorods as intracellular sensor for pH measurement," *J. Appl. Phys.*, vol. 102, pp. 0843048-1–0843048-5, 2007.
[17] Z. Y. Fan and J. G. Lu, "Chemical sensing with ZnO Nanowire FET," *Proc. SPIE*, vol. 6008, pp. 60080H-1–60080H-8, 2005.
[18] W. X. Sun and H. S. Kwok, "Optical properties of epitaxially grown ZnO film on sapphire by pulse laser deposition," *J. Appl. Phys.*, vol. 86, pp. 408–411, 1999.
[19] J. X. Wang, X. W. Sun, A. Wei, Y. Lei, X. P. Cai, C. M. Li, and Z. L. Dong, "ZnO nano-comb biosensor for glucose detection," *Appl. Phys. Lett.*, vol. 88, pp. 233106-1–233106-3, 2006.
[20] A. Wei, X. W. Sun, J. X. Wang, Y. Lei, X. P. Cai, C. M. Li, Z. L. Dong, and W. Huang, "Enzymatic glucose biosensor based on ZnO nanorods array grown by hydrothermal decomposition," *Appl. Phys. Lett.*, vol. 89, pp. 123902-1–123902-3, 2006.
[21] A. Offenhäuser and W. Knoll, "Cell-transistor hybrid systems and their potential applications," *Trends Biotechnol.*, vol. 19, pp. 62–66, 2001.
[22] B. S. Kang, H. T. Wang, F. Ren, S. J. Pearton, T. E. Morey, D. M. Dennis, J. W. Johnsons, P. Rajagopal, J. C. Roberts, E. L. Piner, and K. J. Linthicum, "Enzymic glucose detection using ZnO nanorods on the gate of AlGaN/GaN high electron mobility transistors," *Appl. Phys. Lett.*, vol. 91, pp. 252103-1–252103-3, 2007.
[23] M. Vafaee and H. Youbashizade, "Production of zinc oxide nano-particles by liquid phase processing: An investigation of optical properties," *Mater. Sci. Forum*, vol. 553, pp. 252–256, 2007.
[24] H. Zhang, D. Yang, S. Li, X. Ma, Y. Ji, J. Xu, and D. Qu, "Controllable growth of ZnO nanostructures by citric acid assisted hydrothermal process," *Mater. Lett.*, vol. 59, pp. 1696–1700, 2005.
[25] Z. Zhaochun, H. Baibiao, Y. Yongqin, and C. Deliang, "Electrical properties and raman spectra of undoped and Al-doped ZnO thin films by metalorganic vapor phase epitaxy," *Mater. Sci. Eng. B*, vol. 86, pp. 109–112, 2001.
[26] G. W. Shaw, D. J. Clarement, J. C. Pickup, and P. Bergveld, "Highly sensitive glucose sensor based on work function changes measured by an EMOSFET," *Analyst*, vol. 128, pp. 1062–1066, 2003.
[27] A. Heller, "Electrical connection of enzyme redox centers to electrodes," *J. Phys. Chem.*, vol. 96, pp. 3579–3587, 1992.
[28] Y. Yang, H. Yang, M. Yang, Y. Liu, G. Shen, and R. Yu, "Amperometric glucose biosensor based on surface treated nanoporous ZrO_2/Chitosan composite film as immobilization matrix," *Anal. Chim. Acta.*, vol. 525, pp. 213–220, 2004.
[29] S. Hrapovic, Y. Liu, K. B. Male, and J. H. T. Luong, "Electrochemical biosensing platform using platinum nano-particles and carbon nanotubes," *Anal. Chem.*, vol. 76, pp. 1083–1088, 2004.
[30] P. D. Gaikwad, D. J. Shirale, P. A. Savale, K. Datta, P. Ghosh, A. J. Pathan, G. Rabbani, and M. D. Shirsat, "Development of PANI- PVS-GOD electrode by potentiometric method for determination of glucose," *Int. J. Electrochem. Sci.*, vol. 2, pp. 488–497, 2007.
[31] C.-W. Liao, J.-C. Chou, T.-P. Sun, S.-K. Hsiung, and J.-H. Hsieh, "Preliminary investigations on a glucose biosensor based on the potentiometric principle," *Sens. Actuators B*, vol. 123, pp. 720–726, 2007.

Syed Muhammad Usman Ali received the B.E. degree in electronic engineering from Dawood College of Engineering and Technology (DCET), NED University of Engineering and Technology, Karachi, Pakistan, in 1993, and the M.Sc. degree (Electrical Engineering) in power electronics and computer systems from NED University of Engineering and Technology in 2000. He is currently working toward the Ph.D. degree at the Department of Science and Technology, Linköping University, Linköping, Sweden.

His current research interests include biosensors, microfabrications, nanoelectronics, and nanotechnology.

Omer Nur received the B.Sc. degree (Hons.) in physics from the University of Khartoum, Khartoum, Sudan, in 1986, and the Ph.D. degree in device physics from the University of Linköping, Linköping, Sweden, in 1996.

He is currently an Associate Professor and a Senior Lecturer in the Department of Science and Technology, Linköping University. His current research interests include device physics and technology, and synthesis, characterization, and device development based on ZnO nanostructures for technical and medical applications. He has authored or coauthored more than 120 articles in international journal and in reviewed conference proceedings.

Magnus Willander received the M.Sc. (exp. phys., theo. phys., math. and math.sat.) Fil.Kand degree in 1973 from Lund University, Lund, Sweden, the second M.Sc. (tech. phys., electro phys.) degree in 1974 from Uppsala University, Uppsala, Sweden, and the third M.Sc. (economics) Fil.Kand degree in 1977 from Stockholm.

He is currently engaged in the fundamental problems and applications of solid and soft materials. Since 1995, he has been with Göteborg University, where he was a Full Professor in physics, nanophysics, and mesoscopic physics, and is currently a Guest Professor. Since 2005, he has also been a Full Professor in physical electronics at Linköping University, Linköping, Sweden.

Bengt Danielsson received the Ph.D. degree in biochemistry from Lund University, Lund, Sweeden, in 1979.

Since 1982, he has been an Associate Professor (docent) in biochemistry at Lund University. He was engaged in realizing various biosensor developments, such as the "enzyme thermistor" and "enzyme transistors." He is currently involved in nanotechnology (e.g., ZnO nanowires), bioaffinity arrays, and micropattern formation studied by surface plasmon resonance, ellipsometry, scanning probe microscopy, and chemiluminescent and fluorescent immuno- and molecular imprinting assays. He has authored or coauthored more than 200 publications. His current research interests include bioanalysis and biosensor development, and practical biomedical and environmental applications including miniaturized sensor chips for home and *in* and *ex vivo* monitoring.