Design of Novel Glucose Sensor with In-built Memory Functionality for Real-Time Health Condition Monitoring

Purbasha Ray, Swarnav Mukhopadhyay, and Arpan Deyasi

Abstract A novel glucometer circuit is proposed in this manuscript with an in-built memory chip so that data of a patient can be continuously monitored for the level of glucose present in h(is/er) blood for the last 6 months. The potentiometric Si-based glucose sensor has the added advantage in terms of fabrication point-of-view owing to existing matured microelectronics technology, making it a low-cost device. With 66.3% of sensitivity and 726% higher ON current than the CNT-based glucose sensor as per available published data, the present system may exhibit better acceptability with matured fabrication techniques. Anyone can calculate the percentage changes with the reference level of glucose level, and therefore, we are able to find out the changes with respect to the previous data values. With an easy quantitative analysis, this prototype can easily be accessed by most of the people around the world irrespective of financial limitations.

Keywords Sensitivity · Glucometer · ON current · Continuous tracking · Memory element · Si-technology

1 Introduction

Nowadays, biosensors are the most demanding application we need for our society as daily usage, so the need of manufacturing efficient sensor devices also draws a lot of attention. In recent data, it is said that by 2030, the global population of diabetic patients will reach up to 500 million (Wang 2008). As a result, millions of people are required to test their glucose concentration on blood, and thus, glucose becomes the most tested analyzer in biosensors. Such a high demanded device thus covering 85% of the entire biosensor market (Wang 2008).

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Many researchers recently focused on CNT-based sensors which is not a matured technology as many of our researchers are working still to understand its property (Pitroda et al. 2016). It is quite expensive to fabricate and also not robust. CNT-based sensors are not as compatible with circuits and are not as reliable as Si-based FET biosensor. Now as using CuO nanowire-based extended-gate field-effect transistor (FET) the sensitivity of the sensor can effectively calculate the glucose concentration within 1–12 mM range (Mishra et al. 2020). Whereas theoretically Si-based FET biosensor can go through a more dynamic range, i.e., from 2 to 50 mM glucose concentration which covers the normal diabetic range to a moderate range up to the hyper diabetic range.

Glucometer is a small, low-cost, portable electronic device used to measure the concentration level of glucose in blood. A small test strip is in general attached (Feldman 2003) with this portable device in order to determine the glucose level in the blood of the sample. A few literatures are available where several types of controllers are proposed (Latha et al. 2012; Renesas V850 2011; Gupta and Aggarwal 2013) for efficient and quick data analysis in order to produce low-cost devices. Generally, diabetes is a long-term disease, so its value has to be recorded or observed over months to know its growth, but in previous meters (Latha et al. 2012; Renesas V850 2011) are focused on that certain value when it is taken; and therefore it becomes virtually impossible to get the previous values from the meter as there was no memory element present. Also it is impossible to know the food effect which changes the glucose concentration on blood through the previous glucometer.

In this paper, it has been introduced a potentiometric Si-based FET biosensor. Since Silicon technology is a mature one, more robust and reliable than any other material, so it can be easily considered for fabrication of a new prototype. It can be manufactured and designed in IC technology. Si-based MOSFET sensors are cost-effective too. Amperometric sensors, which are widely used in the market, are very much sensitive to environmental changes and foreign particles, whereas using potentiometric sensors, the sensitivity due to environmental changes is very much less compared to amperometric sensors. Here, Glucose oxide-based enzyme is used to monitor the glucose level in blood, because of its ability to find the glucose analyte accurately. We have performed analytical modeling of the current with respect to glucose concentration for Si-based FET biosensor. So, Si-based FET biosensors have high selectivity, high drain current, accurate response and it is economically cheap.

The following sections include the structural analysis, working principle of the glucose sensor, results and their following discussions. Finally, the paper is concluded with all the advantages of using the Si-based BIO-FET sensor for our society.

2 Structure and Working Principle

As shown in Fig. 1, silicon P-type substrate along with N+ Drain and source region is considered for the design. GO_x is coated above the drain and source electrode as well

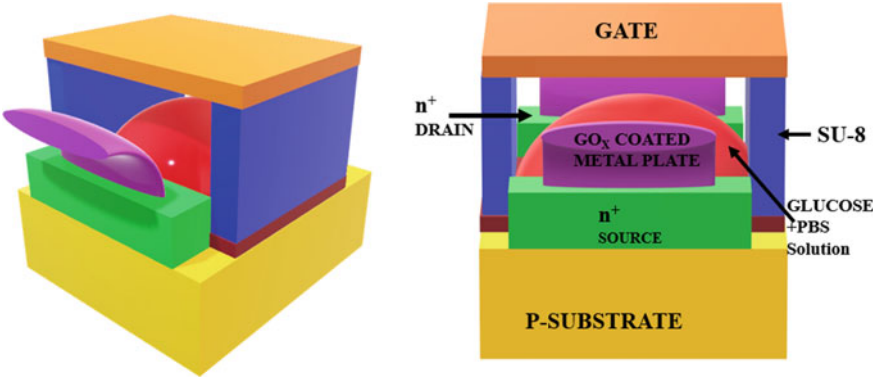


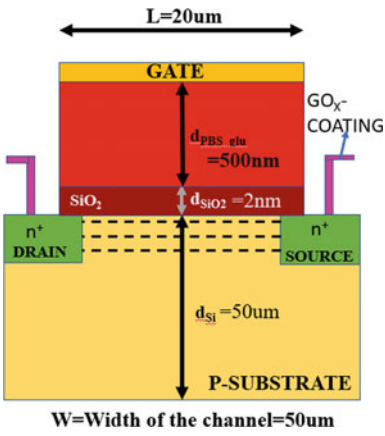
Fig. 1 Basic device structure for the prototype design

as inside the SU-8 walls as the glucose can react in the presence of GO_x with oxygen (O_2) and form gluconic acid ($C_6H_{12}O_7$) and H_2O_2. SiO_2 insulating layer is etched by reactive ion etching to form the cavity. Now the gate electrode is deposited above the SU-8 layer. SU-8 is grown above the SiO_2 layer, which will act as a receptor as well as an insulating layer to protect the substrate from the glucose and PBS solution.

The device length is 20 μm and the width of the device is 50 μm as shown in Fig. 2. The thickness of Cavity, SiO_2 and P-substrate is the following 500 nm, 2 nm and 50 μm, respectively. As the SiO_2 insulating layer is present between substrate and PBS-glucose solution it will act as a capacitor.

As shown in Fig. 3, SiO_2 grown on Si-substrate, it would be in the form of silicon aldehyde where OH group will be present by default at the surface, and thus, it will act as a receptor. When oxide grows over silicon substrate, it has the dangling bonds with oxygen which due to entropy automatically creates a bond with H2-present in

Fig. 2 Dimension of the MOSFET



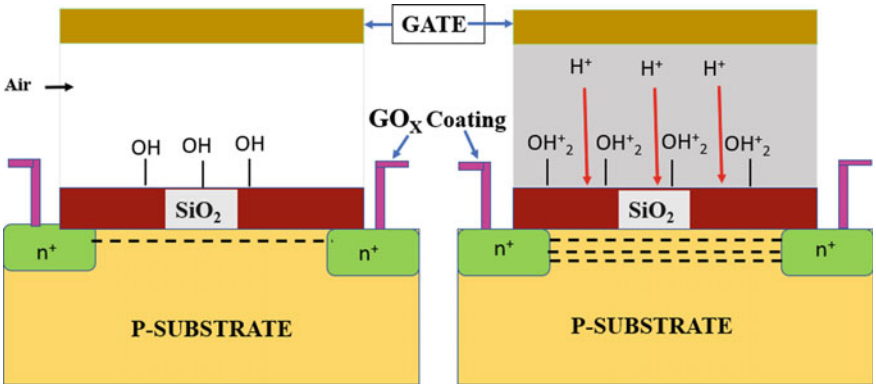


Fig. 3 Ionic movement inside the device

the moisture. So, in the presence of Air, applied gate bias will invert the channel thus producing the drain current.

By applying the PBS solution, due to protonation and deprotonation, charges will be developed above the insulating layer, which will increase the surface potential for which the drain current will eventually increase.

$$\mathbf{SiOH + H^+ \leftrightarrow SiOH_2^+}$$

In the presence of O_2 and GO_x, the glucose solution will create gluconic acid ($C_6H_{12}0_7$) and H_2O_2. H_2O_2 will be ionized and form H^+ and OH^- ions.

$$\mathbf{C_6H_{12}O_6 + O_2 \xrightarrow[GO_x]{} C_6H_{12}O_7 + H_2O_2}$$

Now the H^+ will be directed toward the insulating layer, due to the presence of positive gate bias, hence the charges above the insulating layer will increase, and this will attract a greater number of electrons into the channel and thus increasing the drain current further. Therefore, there will be a net increase in drain current in presence of glucose molecules. OH^- ions will be attracted toward gate electrodes.

3 Analytical Model

According to Si-based FET biosensor, some modeling approaches applied (Rahmani et al. 2013) to the device. Varying the drain voltage (V_d) from 0 to 0.7 V. So, when there is no solution only the air is present in the cavity, then the total capacitance C_{t_AIR} represented below:

$$C_{t\_\mathrm{AIR}} = \frac{C_1 \,.\, C_2 \,.\, C_{\mathrm{AIR}} \,.}{C_1 \,.\, C_2 + C_2 \,.\, C_{\mathrm{AIR}} + C_1 \,.\, C_{\mathrm{AIR}}} \tag{1}$$

where C_1 is the capacitance of substrate region (F/A), C_2 is the capacitance of oxide region (F/A), CAIR is Capacitance of AIR region (F/A)

$$C_{\mathrm{AIR}} = \frac{\xi_{\mathrm{air}}}{d_{\mathrm{cavity}}}; C_1 = \frac{\xi_{\mathrm{Si}}}{d_{\mathrm{Si}}}; C_2 = \frac{\xi_{\mathrm{SiO_2}}}{d_{\mathrm{SiO_2}}}$$

Here, the thickness of d_{cavity} and $d_{\mathrm{PBS\_gluare}}$ same. Now when the PBS-glu will be present in the cavity then capacitance will be

$$C_t = \frac{C_1 C_2 C_3}{C_1 C_2 + C_2 C_3 + C_1 C_3} \tag{2}$$

where C_3 is the capacitance of PBS and glucose region, given by

$$C_3 = \frac{\xi_{\mathrm{glu}}}{d_{\mathrm{PBS\_glu}}}$$

Now the current when there is no solution present in the cavity, drain current is the function of gate voltage (V_{g}) and drain voltage (V_{d}). Thus, the drain current is (Pourasl et al. 2014)

$$I_{D-\mathrm{AIR}} = \phi_{\mathrm{AIR}} \,.\, \frac{2 \,.\, V_{\mathrm{gt}} V_{\mathrm{d}} - V_{\mathrm{d}}^2}{1 + (V_{\mathrm{d}}/V_{\mathrm{c}})} \tag{3}$$

where

$$\phi_{\mathrm{AIR}} = (\mu \,.\, C_{t\_\mathrm{AIR}})/(2 \,.\, L)$$

$$V_{\mathrm{c}} = (V_{\mathrm{sat}}/\mu) \,.\, L$$

'L' indicates effective length of the channel, μ denotes mobility of the carrier, and other parameters have usual meaning.

To analysis, the glucose concentration effect on current, first apply the PBS solution to observe the change in current. Hence, analysis of the I–V curve (Pourasl et al. 2014)

$$I_{D-\mathrm{PBS}} = \phi \,.\, \frac{2 \,.\, (V_{\mathrm{gs}} + V_{\mathrm{pbs}} - V_{\mathrm{t}}) V_{\mathrm{d}} - V_{\mathrm{d}}^2}{1 + (V_{\mathrm{d}}/V_{\mathrm{c}})} \tag{4}$$

where

$$\phi = (\mu \,.\, C_t)/(2 \,.\, L)$$

Now by applying the glucose to the solution in the presence of GO_x and O_2, it will form Gluconic acid and H_2O_2. The corresponding drain current is (Feldman 2003)

$$I_{\mathrm{D-glu}} = \phi \,.\, \frac{2\,.\,(V_{\mathrm{gs}} + V_{\mathrm{pbs}} + V_{\mathrm{glu}} - V_{\mathrm{t}})V_{\mathrm{d}} - V_{\mathrm{d}}^2}{1 + (V_{\mathrm{d}}/V_{\mathrm{c}})} \tag{5}$$

where V_{glu} is the voltage due to glucose concentration. Based on an iteration method demonstrated in []. The concentration control parameter as a function of glucose concentration expressed in a piecewise exponential model as

$$V_{\mathrm{glu\_Fg}} = 1.42 - e^{(-0.1.F_{\mathrm{b}})};\ F_{\mathrm{b}} > 0\,\mathrm{mM} \tag{6}$$

4 Result and Discussion

Based on the equations, we have simulated the behavior of the device and compared it with data of existing costly devices. From Fig. 4 the drain voltage is varying from 0.1 to 0.7 V. When the gate voltage is applied, the surface potential increases, which in turn, enhances the current. But applying the PBS solution, the surface potential increases due to protonation and deprotonation, and thus, surface charge on the SiO_2 layer increases somewhat, due to that more band bending in the channel region so the drain current increases, in presence of PBS solution which is shown in the plot.

From Fig. 5, we can conclude that as the glucose concentration increases, the corresponding drain current eventually increases up to a certain limit, then it saturates.

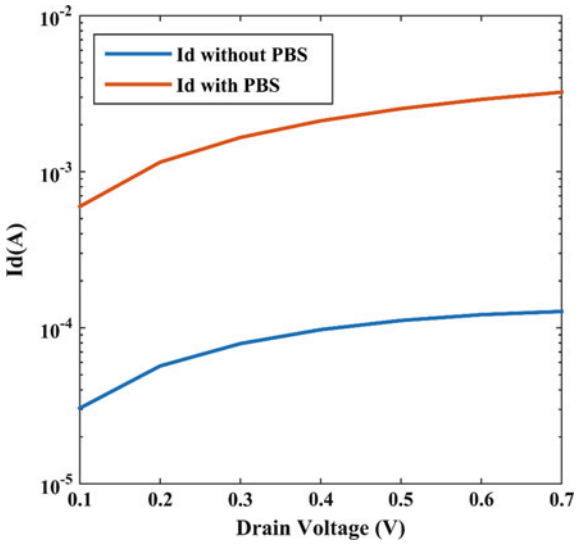


Fig. 4 Variation of drain current with and without PBS

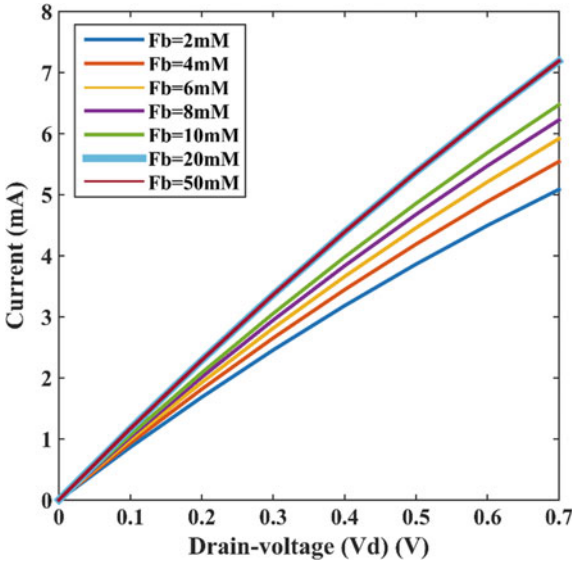


Fig. 5 Variation of drain current for different glucose concentration (Fb)

As concentration of glucose is varied from 2 to 20 mM, the consecutive changes in drain current can be observed very accurately. Above 20 mM concentration, the drain current saturates. Because after reaching a threshold of concentration value, electrons in the channel reach a degeneracy limit.

As shown from Fig. 6, it is quite evident that the sensitivity of Si-based FET BIOsensor is higher than CNT-based sensor. The glucose concentration is varied from 2 to 50 mM and the corresponding sensitivity is calculated accordingly. The sensitivity increases exponentially as the concentration increases. The relative permittivity of Silicon is 11.8 where the relative permittivity of CNT is 3.3 (Pourasl et al. 2014). As the permittivity of Si is more so the net band bending in presence of same potential will be much higher than CNT. Thus, the change in drain current will also be more in Si-based glucose sensors. Thus, the enhancement of sensitivity is around 66.3% compared to CNT-based (Pourasl et al. 2014).

5 Circuit Modeling

In the circuit analysis (Fig. 7), the input current range varies from 5.265 to 7.154 mA, and the input current directly goes to the differential amplifier, which increases the gain, here the gain almost increases up to 10 times. Then the analog output value from the differential amplifier directly goes to Arduino-Uno analog input (here A0). Now the Arduino-Uno is used as an analog-to-digital converter from where we get

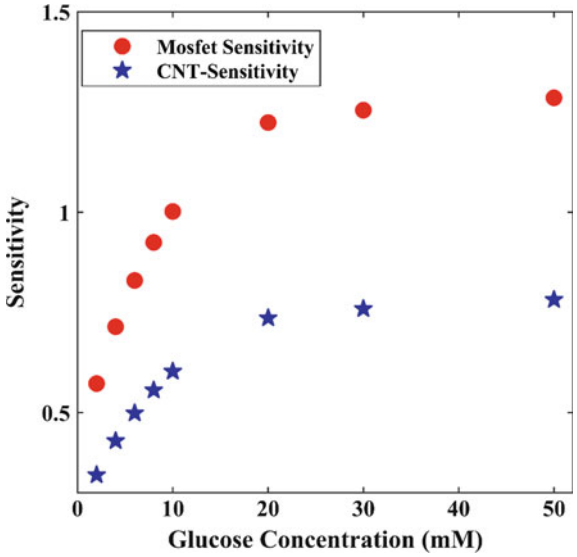


Fig. 6 Comparative analysis of sensitivity with published result

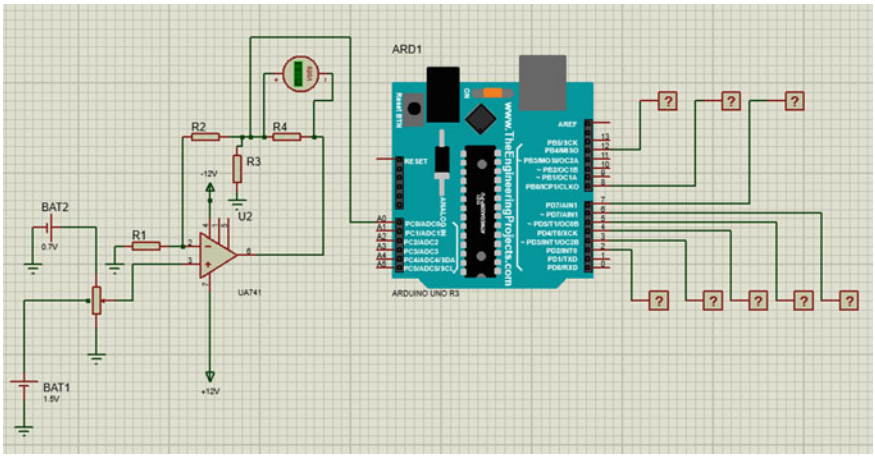


Fig. 7 Circuit layout of the prototype

the digital output as shown in the table. By analyzing the digital output value, the glucose concentration on blood can easily be found.

Corresponding output is represented in Table 1. Percentage change is given in Table 2.

Table 1 Variation of glucose concentration with current

Current (mA)	Digital output								Glucose (mg/dl)
	D7	D6	D5	D4	D3	D2	D1	D0	
5.265	0	0	0	0	0	0	0	0	50
5.39	0	0	0	1	0	0	0	0	60
5.509	0	0	1	0	0	0	0	1	70
5.621	0	0	1	1	0	0	0	0	80
5.727	0	0	1	1	0	0	0	0	90
5.827	**0**	1	0	0	1	1	0	0	100
5.922	0	1	0	1	1	0	0	1	110
6.012	0	1	1	0	0	1	0	1	120
6.097	0	1	1	1	0	0	0	0	130
6.178	0	1	1	1	1	0	1	1	140
6.254	1	0	0	0	0	1	1	0	150
6.326	1	0	0	0	1	1	1	1	160
6.395	1	0	0	1	1	0	0	1	170
6.459	1	0	1	0	0	0	0	1	180
6.521	1	0	1	0	1	0	1	0	190
6.578	1	0	1	1	0	0	1	0	200
6.633	1	0	1	1	1	0	0	1	210
6.685	1	1	0	0	0	0	0	0	220
6.734	1	1	0	0	0	1	1	0	230
6.781	1	1	0	0	1	1	0	1	240
6.825	1	1	0	1	0	0	1	1	250
6.866	1	1	0	1	1	0	0	1	260
6.906	1	1	0	1	1	1	1	0	270
6.943	1	1	1	0	0	0	1	1	280
6.979	1	1	1	0	1	0	0	0	290
7.012	1	1	1	0	1	1	0	1	300
7.044	1	1	1	1	0	0	0	1	310
7.074	1	1	1	1	0	1	0	1	320
7.102	1	1	1	1	1	0	0	1	330
7.129	1	1	1	1	1	1	0	0	340
7.154	1	1	1	1	1	1	1	1	350

Table 2 Computation of percentage change w.r.t. normal dataset

Sl. No	Test name	Normal range (Ref) (mg/dL)	Value (mg/dL)	Percentage change with Ref	Percentage change w.r.t. previous value
1	Glucose Fasting	70–100	105	5% (Increased)	0
	Glucose PP	75–140	125	10.71 (Decreased)	0
2	Glucose Fasting	70–100	113	13% (Increased)	7% (Increased)
	Glucose PP	75–140	168	20.71% (Increased)	30.71% (Increased)
3	Glucose Fasting	70–100	96	4% (Decreased)	3% (Decreased)
	Glucose PP	75–140	130	7% (Decreased)	23.71% (Decreased)
4	Glucose Fasting	70–100	95	5% (Decreased)	2% (Decreased)
	Glucose PP	75–140	130	7% (Decreased)	0
5	Glucose Fasting	70–100	116	16% (Increased)	14% (Increased)
	Glucose PP	75–140	160	14.28% (Increased)	23.07% (Increased)

6 Conclusion

Results obtained from the prototype clearly indicate that the proposed device has improved sensitivity compared to the existing published data, and therefore can be considered as a suitable candidate for glucose level tracking. As the design considers conventional Si-based MOSFET compared to the costly CNT, so the cost of the device may be acceptable for most of the people. The device is extremely suitable for diabetic patients as it may be purchased within financial limits by a large section of the people.

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