Design and simulation of InP and silicon nanowires with different channel characteristic as biosensors to improve output sensitivity

Shahram Mohammadnejad1 · Iman Tahi1 · Mahdiyar Nouri Rezaie1

Received: 21 April 2021 / Accepted: 24 December 2021 / Published online: 9 January 2022
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Abstract
This research contains a comparison among technologies of SiNW-FET/InPNW-FET depending on the size of channel and dopants in channel for biosensing application for two types of silicon and InP materials in the nanowire channel. A device numerical modelling tool, Silvaco ATLAS is used in step one to design three p-type SiNW-FET/InPNW-FET biosensors with a channel width of 40, 60 and 70 nm for these two types of materials and in step two, to design three p-type SiNW-FET/InPNW-FET biosensors with different dopants of $0.1 \times 10^{14}$, $1 \times 10^{14}$ and $10 \times 10^{14}$ cm$^{-3}$ for these two types of materials. Their sensing process is depended on the alteration in charge density which causes changing in the electric field at the surface of the SiNW-FET/InPNW-FET. The resistivity of the device is changed when a negatively charged biomolecules species has a chemical reaction with the external surface of a P-type SiNW-FET/InPNW-FET. To investigate the effect of different channel width and dopants on the performance of the SiNW-FET/InPNW-FET biosensor, several negatively interface charge densities, $Q_F (-0.1 \times 10^{12} \text{,} -0.5 \times 10^{12} \text{ and } -1 \times 10^{12} \text{ cm}^{-2})$ are introduced on the surface of the SiNW-FET/InPNW-FET channel to represent as the actual target analytics (DNA) captured by the bioreceptor of the biosensor. Based on the results, these negatively $Q_F$ attract the hole carriers below the surface of $p$-type nanowire causes to collect carriers in the channel and make an increase in the device output $I_D$. Increment of the applied negative charge density has allowed for more $I_D$ to flow across the channel between drain and source region. The changes of $I_D$ with the applied $Q_F$ are utilized to determine the sensitivities for all designed biosensor with different channel width and channel dopants. The minimum nanowire width of 40 nm with the minimum nanowire dopants of $0.1 \times 10^{14}$ cm$^{-3}$ for the high sensitivity silicon state of 3.6 μA/cm$^{-2}$ compared to the indium phosphide state of 2.8 μA/cm$^{-2}$. Therefore, the best performance for detecting the desired analyte in the silicon state with the lowest width and dopant has been achieved.

Keywords SiNWs · InPNWs · Biosensor · Analyte · SILVACO TCAD
1 Introduction

Nowadays development of electrochemical biosensors in nano devices has been understood in many areas medical investigations (Mathew et al. 2020), environmental checking, industrial testing and etc. A nano biosensor is integrated for detecting a sample by a chemical receptor in contact with a transmitter (Fig. 1a). Recently nanomaterials, like quantum dots, nanowires, carbon nanotubes (CNT), and nano films (Maghsoudi et al. 2021; Singhal et al. 2021), have a good effects on the future of nano biosensors. For instance, the dimension of nanowires of 1-10 nm contains a good property to detect specifications of the most biological entities, like DNA, nucleic, proteins, viruses, bacterium and cells (as illustrated in Fig. 1b).

Nano sensors have been recently one of the most appropriate methods for rapid analysis of body cells (Hashim and Shakib 2020; Christopher et al. 2020). The common chemical detection techniques are insufficient because of its kind of detection and on-site recognition (Christopher et al. 2020). Besides, high accuracy and precision of biological and chemical species recognition and measurement are very crucial in the medical diagnostics and environmental monitoring fields, which can be provided with the utilization of biosensor (Ekrami et al. 2021). To meet this requirement, the exploitation of semiconductor nanostructures as a promising piece to be integrated in biosensing application (Saylan et al. 2019; Campos et al. 2020) has been considered by researchers in the recent years. Among them, nanowire field-effect transistor (NW-FET) has emerged and attracted attention to be applied in designing a biosensor due to their extremely high sensitivity, selectivity, real-time detection, and label-free. The sensing of this kind of transistor is like the other ones when changing in charge density, causes changing in the electric field at the NW-FET surface. For example, biomolecules with a negative charge on the surface of a p-type FET causes changing in the device resistance (Christopher et al. 2020). Because of the large surface-to-volume, electrical properties, and incompatibility, the NW-FET have been a good detector with ultrasensitive and selective property (Chen et al. 2011). The typical FET-based device consists of three electrodes, named as drain, gate, and source. The source and drain electrodes are connected together by a channel made of NW-FET, while the gate

![Fig. 1](image)

(a) Biological recognition element; (b) Dimension of nanowire
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The electrode is responsible for the modulation of channel conductance (Tran et al. 2018). The sensing mechanism of the device is based on the difference in charge density originated from the bounded target analyte at the NW-FET exterior surface and causes the change of electric field at the channel. The resistance of the device may increase or decrease which effect on the electrical conductivity/resistivity of 1-D nanostructures between electrodes (Nuzaihan et al. 2016; Chartuprayoon et al. 2015).

The silicon nanowire FET demonstrates changing in the channel conductance, which is influenced by the external pH. Variation in the surface charge density with the protonation and deprotonation of biological and chemical species results in a variation in the conductance of the device. Patolsky et al. electrically realized the influenza (type A) virus using $p$-type silicon nanowire FET in an array (Patolsky et al. 2004). They could reach the effect of the pH value on the conductance which is influenced by the surface charge density of the virus articles. In this regard, the pH variation is effective to alter the conductance coincided with the binding (or unbinding) of the virus particles. When the conductance is changing with pH around 6.8, it shows that the value of the isoelectric point (IEP) of the target virus is changing between pH of 6.5 and 7.0. The results were in good aligned with the results reached from electrophoretic mobility measurements. Therefore, it illustrated that the nanowire FET is also a useful tool for the determination of isoelectric points of the target virus. Figure 2 displays a schematic illustration of the antibody-modified silicon nanowire FET used for detecting the influenza virus.

Cui et al. (Cui et al. 2001) also evolved a FET biosensor with silicon nanowires. In this kind of device, boron-doped silicon nanowires were utilized as a bridge between the source and the drain of the silicon nanowire FET (SiNW FET). This FET device was designed of a high-quality silicon nanowire put into use with amine and oxide materials for highly sensitive, real-time detection (Cui et al. 2001). After that, researchers investigated that the number of bridging nanowires on nano sensors had an effect on biosensing sensitivity. Zhang et al. (2004) demonstrated that multiple $\text{In}_2\text{O}_3$ nanowire FET devices were more sensitive than single-nanowire devices for gaseous chemical sensing. While the underlying mechanism was not determined, the authors surmised that this achievement might be influenced to the shape...
of nanowire connections in coinciding nanowires. Gruner et al. (2006) further suggested that sensitivity in individual CNT-FET biosensors, was found to be maximal for single-nanowire devices and decreased with increasing number of nanowires indicating that single-nanowire devices can yield the highest sensitivity. On the other hand, researchers focused on many materials such as ZnO. Zhiyong Fan and Jia G. Lu (2006) fabricated nano sensor with Si/SiO₂/ZnO/Al/Ni structure, but the sensitivity for this structure was low. After that, some researchers focused on dimensions of the structure and dopants for optimizing. In this regard, Elfstrom et al. (2007) designed the size-dependent surface charge sensitivity of silicon nanowires, reported that silicon nanowires with different widths display various electrical performance levels of their device sensitivity during biological responses. After a while, M. Fathil et al (2018) designed a nano sensor with Si/SiO₂/Si/Pt/Ni structure which its sensitivity has been changed with different length and as a result, he could get a sensitivity with 2.17 µA/cm² value (Table 2), considering minimum length and fixed parameters (Table 1).

In this paper, NW-FET sensors are simulated based on two structures. At first, SiNW-FET with Si/SiO₂/Si/Al/Co structure and then InPNW-FET with Si/SiO₂/InP/Al/Co structure are simulated. In the following, two designs are considered for both of the structures. First, the channel region with different width of 40, 60 and 70 nm is patterned from the top-Si layer to bridge the drain and source. The length and dopant of the SiNW(InPNW) are kept constant at 1 µm and 1×10¹⁴ cm⁻³ for all channel width designs, respectively. Second, the top-Si layer is doped with p-type dopant at concentration of 0.1×10¹⁴, 1×10¹⁴ and 10×10¹⁴ cm⁻³ and the width of the SiNW(InPNW) is kept constant at 40 nm for all channel dopant designs in order to understand the effect of the channel doping on the performance of the biosensor. The changes on the electrical resistance and I_D due to different channel width and channel doping are observed from the TONYPLOT application in the Silvaco ATLAS. In addition, the surface of the channel is applied with negative Q_F of −0.1×10¹², −0.5×10¹² and −1.0×10¹² cm⁻² as representation of the existence charged target DNA analyte which presents on SiNW-FET (InPNW-FET) biosensor upon detection. The values of I_D are increased when more negative Q_F values are applied onto the SiNW/InPNW channel surface for every simulated device with different channel width and dopants. However, the current generates by the component at a voltage of −1 V in the values of −12, −16 and −17 µA in the silicon state compared with the values of −4, −4.5 and −5 µA in the indium state by changing the resistance with channel widths of 40, 60 and 70 nm. Furthermore, the current generates by the component at a voltage of −1 V in the values of −20, −180 and −1400 µA in the silicon state compared with the values of −1.8, −16 and −75 µA in the indium state by changing the resistance with channel doping of 0.1×10¹⁴, 1×10¹⁴ and 10×10¹⁴ cm⁻³. Moreover, to analyze the effect of different channel widths on the sensitivity of the biosensor, the I_D versus negative Q_F for channel width from 40 to 70 nm with p-type dopant at concentration of 1×10¹⁴ cm⁻³ is plotted. In other simulation, in order to analyze the effect of doping concentration on the sensitivity of the biosensor, the I_D versus negative Q_F for different channel doping is plotted. For both of the SiNW-FET and InPNW- FET, the increasing of sensitivity has been compared in Table 2 with different channel widths and dopants in order to get the best sensitivity.
Table 1 Parameters of Si and InP used in simulation

| Parameters                  | Units       | Si          | InP         | Ref (Fathil, et al. 2018) |
|-----------------------------|-------------|-------------|-------------|---------------------------|
| Length                      | μm          | 1           | 1           | 0.5_10                    |
| Width                       | nm          | 40_70       | 40_70       | 50                        |
| thickness                   | nm          | 44          | 44          |                           |
| Doping                      | cm⁻³        | 0.1×10¹⁴, 1×10¹⁴, 10×10¹⁴ | 1×10¹⁸          | 1×10¹⁸, -0.5×10¹², and -1×10¹² |
| Q factor(Qf)                | cm⁻²        | -0.1×10¹², -0.5×10¹², and -1×10¹² |                          |                           |
| V_D                         | Volt        | -1          | -1          | -1                        |
| I_D                         | μA          | -12_−17     | -4_−5       | 7.28 _ 0.47               |
| σ_Diatl                     | nm          | 1           | 1           |                           |
| σ_BL                        | nm          | 80          | 80          |                           |
| σ_OHP                       | nm          | 0.33        | 0.33        |                           |
| Band gap (Eg)              | eV          | 1.1         | 1.344       | 1.1                       |
| Dielectric constant        | –           | 11.7        | 12.5        | 11.7                      |
| Electron mobility (mn)     | cm²/V s     | 1000        | 5400        | 1000                      |
| Hole mobility (mp)         | cm²/V s     | 500         | 200         | 500                       |
| Effective DOS for electrons (Nc) | cm⁻³  | 2.86E + 19 | 5.70E + 17 | 2.86E + 19 |
| Effective DOS for electrons (Nv) | cm⁻³  | 2.66E + 19 | 1.10E + 19 | 2.66E + 19 |
| Electrodes                 | Units       | Cobalt      | Nickel      |                           |
| Thickness                   | nm          | 100         | 100         |                           |
| Work function (f)          | eV          | 5           | 5.01        |                           |
| Reference                   |             |             | Levinshtein (1997) |                           |
2 Simulation method

2.1 Basic algorithms and methods for structural design

To understand the performance of nanosensors, we should understand relation between the potential at the gate oxide surface and the amount of analyte (PH) in the electrolyte (Manjakkal et al. 2020). An electrical layer (EL) is defined between solid and liquid interfaces. The potential at the gate is shown with $\Psi_s$ and depends on the chemical reactions as in Eq. (1) (Failed 2020; Rani et al. 2018):

$$\Psi_s = \frac{\sigma_s}{C_d}$$

In this relation $\sigma_s$ is the density of surface-charge and $C_d$ is the capacity of EL zone. The reaction between the charged particles with the gate surface oxide changes the voltage ($\Psi_s$), which is in relation with the threshold voltage ($V_{th}$) of the TFT (Choi et al. 2020). In a Si TFT, SiO$_2$ in gate oxid consists of Si–OH as hydroxyl groups are affected by the absorption or repulsion of protons that is affected by the pH value of the electrolyte solution. This absorption or repulsion causes readjust in ionic component of the electrolyte and at last in the constitution of an EL with a certain distance from the gate. The shape of the EL can be illustrated by complicated models like the Gouy-Chapman-Stern theory in Fig. 3. The EL applies surface potential accumulation to the oxide surface, which is then the pH can be sensed with this mechanism. This potential ($\Psi_s$) is in a direct relation with pH value in the electrolyte solution, $\Psi_s = f(pH)$ (Bergveld 2003).

The acid/base reactions happens between the gate oxide surface with hydroxyl groups (Si–OH) and the hydrogen ions ($H^+$) in the electrolyte solution with this model, which is described with $K_a$ and $K_b$ as follows Eqs. (2), (3) (Hal et al. 1996):

$$Si - OH_2^+ \rightleftharpoons Si - OH + H^+, K_b = \frac{[Si - OH][H^+]}{[Si - OH_2^+]},$$

$$Si - OH \rightleftharpoons SiO^+ + H^+, K_a = \frac{[SiO^+][H^+]}{[Si - OH]}.$$

We can see that $[H^+]_s$ is the activity of the ions at the surface and as the surface units per area. The charges from the surface with zero potential in the bulk area of the solution.

| Material | Doping(cm$^{-3}$) | Width (nm) | High sensitivity ($\mu A/cm^2$) |
|----------|-------------------|------------|-------------------------------|
|          | 40                | 60         | 70               | This work | Ref (Fathil, et al. 2018) |
| Si       | $0.1 \times 10^{14}$ | 3.6        | 2.8              | 2.1      | 3.6                | 2.17 |
|          | $1 \times 10^{14}$   | 3.2        | 2.1              | 1.8      | 3.6                | 1.5  |
|          | $10 \times 10^{14}$  | 2.5        | 1.8              | 1.4      | 10.1               | 1.4  |
| InP      | $0.1 \times 10^{14}$ | 2.8        | 1.9              | 1.3      | 2.8                |     |
|          | $1 \times 10^{14}$   | 2.4        | 1.5              | 0.78     |                     |     |
|          | $10 \times 10^{14}$  | 1.8        | 1.1              | 0.65     |                     |     |
form the potential at the solid–liquid zone. This absorption or repulsion of hydroxyl groups in oxide surface causes the H⁺ ion concentration varies at the both of solid–liquid interface and bulk electrolyte solution (Hal et al. 1996). The Boltzmann distribution relates that the gate surface with H⁺ ions ([H⁺]s) and PHs is in relation with the bulk electrolyte with H⁺ ions ([H⁺]b) and PHb by Eqs. (4–7) (Hal et al. 1996):

\[
[H^+]_s = [H^+]_b \exp\left(-\frac{q\Psi_s}{kT}\right) \tag{4}
\]

\[
pH_s = pH_b + \frac{q\Psi_s}{2.3kT} \tag{5}
\]

\[
pH_s = -\log_{10}[H^+]_s \tag{6}
\]

\[
pH_b = -\log_{10}[H^+]_b \tag{7}
\]

We assume q as the primary charge, k as the Boltzmann constant and T as the absolute temperature.

Ns is the total density from level sites at the gate oxide and determine as Eq. (8) (Rani et al. 2018):

\[
N_s = [Si - OH] + [SiO^-] + [Si - OH_2^+] \tag{8}
\]
The parameters $K_a$, $K_b$ and $N_s$ is in relation with the type of gate insulator. $\sigma_s$ is the surface-charge density with the difference in number of positively charged groups and negatively charged groups per unit area as Eq. (9) (Rani et al. 2018):

$$\sigma_s = q([Si - OH]^+ - [SiO^-]) = qN_s \left( \frac{[H^+]^2_s - K_aK_b}{K_aK_b + Kb[H^+]_s + [H^+]^2_s} \right) = -q[B]$$ (9)

The point of zero charge (the number of positively charged groups equals the number of negatively charged) in Eq. (10) is in $pH_{pzc} = 2$ for SiO$_2$ as gate oxide, the point of zero charge is $[B] = \sigma_s = 0$ (Rani et al. 2018).

$$pH_{pzc} = (pK_a + pK_b)/2$$ (10)

Dividing Eq. (9) with $pH_s$ gives Eq. (11) (Hal et al. 1996):

$$\frac{\partial \sigma_s}{\partial pH_s} = -q \frac{\partial [B]}{\partial pH_s} = -q \frac{\partial \beta_{int}}{\partial pH_s}$$ (11)

$\beta_{int}$ measures the chargeability of the oxide surface by changing the pH (Stolze et al. 2020) value of the solution. It is in relation with the intrinsic properties of the surface, $N_s$, $K_a$ and $K_b$. At the EL because of the charge neutralization condition, an equal value of charge density $\sigma_{EL}$ appears with the opposite pole on the electrolyte side of the layer. As you can see in Fig. 3, the EL is formed of several layers (the Stern layer and the diffuse layer (DL)) depending on the distribution of ionic ingredient of the electrolyte (Hal et al. 1996). At a distance of $\times 1$ from the gate oxide adsorbate ions and molecules that is close to the gate oxide is called the inner Helmholtz (IHP) (Fig. 3). The surface-charge density ($\sigma_s$) on the gate oxide which can only approach the oxide level up to a distance of $\times 2$ is balanced by dissolved ions. The outer Helmholtz plane (OHP) is the area between the IHP and the nearest available solvated ions. DL area is between the OHP and the bulk of the electrolyte where the value of the solvated ions decreases. Electrostatic force or higher ionic strength electrolyte causes the increasing of interaction between the surface charge and the solvated ions and form a thinner DL. The potential inside the Stern layer decreases with increasing of distance between the gate oxide surface and the electrolyte solution. In the following the potential into the DL zone decreases until become near zero (Fig. 3, red curve) in the bulk solution ($\Psi_b$). There is another potential, between the unmoved ions of the Stern layer and the moving ions of the DL with $\Psi_2$, (Fig. 3). There are a parameter as Debye length ($\lambda_D$) that illustrate the distance from the oxide surface in the electrolyte solution, in this area the potential decreases from its real value in the gate oxide (Wang et al. 2020). For a given electrolyte solution, $\lambda_D$ can be calculated by Eq. (12) (Rani et al. 2018):

$$\lambda_D = \sqrt{\frac{\varepsilon_0 \varepsilon_r K T}{2 q^2 P}}$$, \hspace{1cm} $P = \frac{1}{2} \sum c_i z_i^2$ (12)

$\varepsilon$ is the dielectric constant of the electrolyte solution, $\varepsilon_0$ is the vacuum permittivity, $P$ is the ionic strength of the electrolyte and the concentration of ions in the electrolyte solution display with $c_i$ and $z_i$. The surface-charge density $\sigma_s$ is in relation with charges in the electrolyte side of EL ($\sigma_{EL}$) (as shown in Fig. 3) as Eq. (13) (Hal et al. 1996):

$$\sigma_s = -\sigma_{EDL} = \sigma_{OHP} + \sigma_{DL}$$ (13)
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σOHP and σDL depict the charge densities in the OHP and DL, respectively (Simulated parameters in Table 1). At last, the charge is in relation with the capacitance in this area($C_{EDL}$) by Eq. (14), Eq. (15) (Hal et al. 1996).

$$\sigma_s \approx C_{EDL} \Psi_s$$  \hspace{1cm} (14)

$$\frac{\partial \sigma_s}{\partial \Psi_s} \approx C_d$$  \hspace{1cm} (15)

With Cd capacitance, combining Eqs. (11) and (15), the dependence of $\Psi_s$ on the pHb can be obtained as Eq. (16) (Hal et al. 1996):

$$\frac{\partial \Psi_s}{\partial \text{pH}_b} = -2.3 \frac{KT}{q} - a, \quad a = \frac{1}{1 + 2.3 \frac{KTC_L}{qF_{\text{int}}}}$$  \hspace{1cm} (16)

We can see that there is a direct relation between the pH sensitivity of the TFT and the charging behavior of oxides. In this model the site-dissociation developed by Yates is used to describe the FET sensitivity (Hal et al. 1996). Therefore because of the relation in charging and dimensional properties we can also understand the relation in the sensitivity with the variation of silicon nanowire (SiNW) width (Wsi) and doping (Choi et al. 2010), and compare them with InP nanowire(InPNW), on the other hand to compare the different effects of channel width and doping with these two materials, as their parameters are near together, the electrical parameters which are used in simulation of these two materials have been declared in Table 1. Also, in this table parameters for a reference are given.

Furthermore, one of the aspects that may influence the sensitivity of the FET device is the wire size, which is highly related to the surface-to-volume ratio. For an example, if charged analyte is applied to a thick wire, only specific area located at or near to the wire surface with the exception to the interior site of the wire will be affected by the applied electric field originated from the charged target analyte (Zhang et al. 2020). On the other hand, when the diameter of the wire is decreased down to nanoscale, the surface-to-volume ratio can be drastically improved and the entire nanowire cross-section area could be influenced by the external electric field and exceptional change in induced conductance can be achieved inside the SiNW-FET/InPNW-FET (Fatihil, et al. 2018; Sadighbayan et al. 2020). The high quality of the SiNW-FET/InPNW-FET is essential to limit the distribution of charge carriers on the sidewalls and, thus enable for high electrical conductance and sensitivity of the NW (Namdari et al. 2016).

As an alternative, two ways that may improve the performance of p-type SiNW-FET/InPNW-FET biosensor is by optimizing the width and dopants of the nanowire (Zafar et al. 2018). In this paper, the effect of different channel width and dopants on electrical characteristic of the SiNW-FET/InPNW-FET biosensor is investigated via simulation in Silvaco ATLAS device modelling tool. In step one SiNW-FET/InPNW-FET biosensors with several channel widths of 40, 60 and 70 nm are designed and simulated and in step two with several channel doping of $0.1 \times 10^{14}$, $1 \times 10^{14}$ and $10 \times 10^{14}$ cm$^{-3}$ are designed and simulated to observe the changes in the electrical resistance and output drain current ($I_D$) of the device and at last both of them combine to achieve the best result. Furthermore, the SiNW-FET/InPNW-FET biosensor is introduced with several negative interface charge densities, $Q_F$ including $-0.1 \times 10^{12}$, $-0.5 \times 10^{12}$ and $-1 \times 10^{12}$ cm$^{-2}$ on the surface of channel to represent target biomolecule (Hashim et al. 2008; Doucey and Carrara 2019) for every simulated channel width/dopants, in order to observe their impact on the I-V characteristic and hole
concentration beneath the surface of the SiNW. Finally, the sensitivities of the SiNW-FET/InPNW-FET biosensors with different channel width and dopants are determined to get the best result (Liu et al. 2020). The result should suggest an appropriate choice of channel width and dopants of SiNW-FET/InPNW-FET for future biomedical application in biosensing.

2.2 Device Simulation

2.2.1 Research methodology

The simulation is started by declaring the structure of the SiNW-FET (InPNW-FET) biosensor through the use of specific ATLAS commands, such as mesh, region, electrode, doping, and etc. Silicon (Si) with low p-type doping concentration of $1 \times 10^{14} \text{ cm}^{-3}$ is utilized as the biosensor’s substrate. A 145 nm-thick silicon dioxide ($\text{SiO}_2$) layer is used as an electrical passivation layer known as buried oxide (BOX) on top of the substrate layer. Next, another Si layer with thickness of 44 nm is deposited on the surface of the BOX to become the top-Si device layer. In the following two designs is considered.

2.2.2 Different channel width

The top-Si layer is doped with p-type dopant at concentration of $1 \times 10^{14} \text{ cm}^{-3}$. It is then patterned in order to form three regions, which are source, drain, and channel of the SiNW-FET (InPNW-FET) biosensor. The source and drain regions are patterned with the length and width of $500 \times 500 \text{ (nm)}^2$, while the channel region with different channel width of 40, 60 and 70 nm is patterned from the top-Si layer to bridge the drain and source, the length of the SiNW(InPNW) is kept constant at 1 μm for all channel width designs (Sang et al. 2016). In order to provide a biomolecule interaction area, another layer of SiO$_2$ with thickness of 15 nm is covered on the surface of the SiNW(InPNW)-FET channel. The surface of drain and source pads are deposited with 40 nm-thick aluminum and 100 nm-thick cobalt layer to provide electrodes for the device.

2.2.3 Different channel doping

The top-Si layer is doped with p-type dopant at concentration of $0.1 \times 10^{14}$, $1 \times 10^{14}$ and $10 \times 10^{14} \text{ cm}^{-3}$ in order to understand the effect of the channel doping on the performance of the biosensor. It is then patterned in order to form three regions, which are source, drain, and channel of the SiNW-FET (InPNW-FET) biosensor. The source and drain regions are patterned with the length and width of $500 \times 500 \text{ (nm)}^2$, while the channel region width of 40 nm is patterned, the length of the SiNW(InPNW) is kept constant at 1 μm for all channel doping designs (Sang et al. 2016). In order to provide a biomolecule interaction area, another layer of SiO$_2$ with thickness of 15 nm is covered on the surface of the SiNW(InPNW)-FET channel. The surface of drain and source pads are deposited with 40 nm-thick aluminum and 100 nm-thick cobalt layer to provide electrodes for the device. Fig. 4a demonstrates the cross-sectional structure views of the SiNW-FET biosensor is designed in Silvaco ATLAS. In the next step, instead of using silicon in the relevant channel, indium phosphide is used as InPNW-FET, which can be seen in Fig. 4b.
2.2.4 Device modelling of the SiNW-FET and InPNW-FET biosensor

In first step the investigation of the effect of different channel width on the electrical characteristic of the SiNW-FET(InPNW-FET) biosensor is performed by varying the width of the SiNW(InPNW-FET) at 40, 60 and 70 nm and in the second step the investigation of the effect of different channel doping on the electrical characteristic of the SiNW-FET(InPNW-FET) biosensor is performed by varying doping of the SiNW (InPNW-FET) at 0.1×10^{14}, 1×10^{14} and 10×10^{14} cm^{-3}. The changes on the electrical resistance and \( I_D \) due to different channel width and channel doping are observed from the TONYPLOT application in the Silvaco ATLAS. In addition, the surface of the channel is applied with negative \( Q_F \) of \(-0.1\times10^{12}, -0.5\times10^{12} \) and \(-1.0\times10^{12} \) cm\(^{-2}\) as representation of the existence of charged target DNA analyte that presents on SiNW-FET (InPNW-FET) biosensor upon detection (Lu et al. 2016). The applied \( Q_F \) has an effect on the electrical characteristic of current-voltage (I–V) and hole concentration for different channel width and channel doping of the SiNW-FET (InPNW-FET) biosensor. When \( I_D \) changes we can determine the sensitivity of the device. So we can understand the performance of the biosensors with different channel width and channel doping to produce it for the future (Puppo et al. 2016).

3 Results and discussion

3.1 Channel width effect

3.1.1 Channel width effect on electrical characteristic

The I–V characteristics of the SiNW-FET and InPNW-FET biosensor for different channel width are displayed as in Fig. 5a and b. However, the current generates by the component at a voltage of \(-1 \) V in the values of \(-12, -16 \) and \(-17 \) µA in the silicon state and the values of \(-4, -4.5 \) and \(-5 \) µA in the indium state by changing the resistance with channel widths of 40, 60 and 70 nm. The increasing of \( I_D \) with the increase of the channel width of the device is well-correlated with the Eq. (17) which describe the relationship between

![Fig. 4 SiNW-FET biosensor simulated in Silvaco ATLAS TonyPlot 2D cross-sectional views along the SiNW- FET; b 2D cross-sectional views along the InPNW- FET](image-url)
electrical resistance, resistivity, and dimension of the SiNW (InPNW-FET) channel (Fathil et al. 2018):

\[ R = \rho \cdot \frac{L}{WT} \]  \hspace{1cm} (17)

where \( R \) is the electrical resistance, \( \rho \) is the resistivity, \( L \) is the length, \( W \) is the width, and \( T \) is the thickness of the SiNW (InPNW-FET) channel. From the Eq. (17), it can be seen that the \( R \) is indirectly proportional to the \( W \) of the SiNW (InPNW-FET) channel where the \( R \) is decreased with the increasing of SiNW (InPNW-FET) channel width of 40, 60 and 70 nm, respectively (Elfström et al. 2007). Since the Ohm’s law \( I_D \) versus \( V_D \) characteristic and stated that the current (I) is indirectly proportional to the \( R \), the increase of channel width has increased the output \( I_D \) as shown in Fig. 5a and b.

### 3.1.2 Application of interface charge densities for difference channel widths

The effect due to the application of different \( Q_F \) values on the channel surface of the SiNW-FET and InPNW-FET biosensor for different channel width of 40, 60 and 70 nm towards the \( I_D \) versus \( V_D \) characteristics is shown as in Fig. 6a–f (Reddy and Panda 2020). The surface of channel is applied with negative \( Q_F \) of \(-0.1 \times 10^{12}, -0.5 \times 10^{12}\) and \(-1 \times 10^{12}\) cm\(^{-2}\) to mimic the different concentrations of target biomolecules (Hashim et al. 2008). The values of \( I_D \) are increased when more negative \( Q_F \) values are applied onto the SiNW channel surface for every simulated device with different channel width as shown in Fig. 6a–c and onto the InPNW channel surface for every simulated device with different channel width as shown in Fig. 6d, e and f. These changes in \( I_D \) are due to the fact that the SiNW (InPNW) channel is doped with p-type dopant with doping concentration of \( 1 \times 10^{14}\) cm\(^{-3}\), which comprise of hole as majority carrier. In other word, the channel of the device is mainly comprised of positive charge.

More negative \( Q_F \) on the SiNW (InPNW) channel surface has attracted the hole carriers underneath the channel surface and contribute to a good formation of hole conduction channel across the SiNW (InPNW). Therefore, more \( I_D \) is allowed to flow from drain to source. These results illustrates that all the simulated SiNW (InPNW) channel width is affected by the application of different \( Q_F \) values applied on them, hence indicate that the SiNW (InPNW) channel detect different target analyte concentrations that will be captured by the bioreceptor immobilized onto the device. The detection will be signified by

Fig. 5 Electrical characteristics of the a SiNW-FET; b InPNW biosensor simulated in Silvaco ATLAS due to the effect of different channel width of 40 nm, 60 nm, and 70 nm.
the relative change in $I_D$, which is the percentage difference of $I_D$ before and after detection (Reddy and Panda 2020).

3.2 Channel doping effect

3.2.1 Channel doping effect on electrical characteristic

The I–V characteristics of the SiNW-FET and InPNW-FET biosensor for different channel doping are displayed as in Fig. 7a and b. However, the current generates by the component at a voltage of $-1$ V in the values of $-20$, $-180$ and $-1400$ $\mu$A in the silicon state and the
values of −1.8, −16 and −75 µA in the indium state by changing the resistance with channel doping of 0.1×10^{14}, 1×10^{14} and 10×10^{14} cm^{-3}. The increasing of $I_D$ with the decreasing $p$ because of the increasing in the channel doping of the device is well-correlated with the Eq. (18) which describe the relationship between electrical resistance, resistivity, and dimension of the SiNW (InPNW-FET) channel (Fathil, et al. 2018):

$$R = \rho \cdot \frac{L}{WT}$$  \hspace{1cm} (18)

where $R$ is the electrical resistance, $\rho$ is the resistivity, $L$ is the length, $W$ is the width, and $T$ is the thickness of the SiNW (InPNW-FET) channel. From the Eq. (18), it can be seen that the $\rho$ is directly proportional to the $R$ of the SiNW (InPNW-FET) channel and because of inverse relation between $\rho$ and doping (Li et al. 2011), the $R$ is decreased with the increasing of SiNW (InPNW-FET) channel doping of $0.1 \times 10^{14}$, $1 \times 10^{14}$ and $10 \times 10^{14}$ cm$^{-3}$, respectively. Since the Ohm’s law $I_D$ versus $V_D$ characteristic and stated that the current ($I$) is indirectly proportional to the $R$, the increase of channel doping has increased the output $I_D$ as shown in Fig. 7a and b.

### 3.2.2 Application of interface charge densities for difference channel doping

The effect due to the application of different $Q_F$ values on the channel surface of the SiNW-FET and InPNW-FET biosensor for different channel doping of $0.1 \times 10^{14}$, $1 \times 10^{14}$ and $10 \times 10^{14}$ cm$^{-3}$ towards the $I_D$ versus $V_D$ characteristics is shown as in Fig. 8a–f (Reddy and Panda 2020). The surface of channel is applied with negative $Q_F$ of $-0.1 \times 10^{12}$, $-0.5 \times 10^{12}$ and $-1 \times 10^{12}$ cm$^{-2}$ to mimic the different concentrations of target biomolecules (Hashim et al. 2008). The values of $I_D$ are increased when more negative $Q_F$ values are applied onto the SiNW channel surface for every simulated device with different channel doping as shown in Fig. 8a–c and onto the InPNW channel surface for every simulated device with different channel doping as shown in Figs. 8d, e and f. These changes in $I_D$ are due to the fact that the SiNW (InPNW) channel is doped with p-type dopant, which comprise of hole as majority carrier. In other word, the channel of the device is mainly comprised of positive charge.

More negative $Q_F$ on the SiNW (InPNW) channel surface has attracted the hole carriers underneath the channel surface and contribute to a good formation of hole...
conduction channel across the SiNW (InPNW). Therefore, more $I_D$ is allowed to flow from drain to source. These results illustrate that all the simulated SiNW (InPNW) channel doping is affected by the application of different $Q_F$ values applied on them, hence indicate that the SiNW (InPNW) channel detect different target analyte concentrations that will be captured by the bioreceptor immobilized onto the device. The detection will

![Graphs showing electrical characteristics](image-url)

**Fig. 8** Electrical characteristics in terms $I_D$ versus $V_D$ characteristics (a, b, c) SiNW-FET; (d, e, f) InPNW-FET graph for biosensors at different channel doping of $0.1 \times 10^{14}$ cm$^{-3}$, $1 \times 10^{14}$ cm$^{-3}$, $10 \times 10^{14}$ cm$^{-3}$ when applied with different $Q_F$ of $-0.1 \times 10^{12}$ cm$^{-2}$, $-0.5 \times 10^{12}$ cm$^{-2}$, and $-1.0 \times 10^{12}$ cm$^{-2}$
be signified by the relative change in $I_D$, which is the percentage difference of $I_D$ before and after detection (Reddy and Panda 2020).

3.3 Device performance analysis: sensitivity

The accuracy of the measured output signal for a set of measurement with various analyte concentration through a straight-line equation is a feature that can be utilized to determine the sensitivity of a biosensor from a mathematic representation as in Eq. (19) (Fathil, et al. 2018):

$$y = mx + c$$  \hspace{1cm} (19)

where $c$ is the analyte concentration, $y$ is the output feedback, and $m$ is the biosensor’s sensitivity (Fan and Lu 2006). The device’s sensitivity can be determined through calculation based on the ratio of difference of output $I_D$ ($\Delta I_D$) to difference of applied negative $Q_F$ ($\Delta Q_F$) on the SiNW-FET/InPNW-FET surface for the three channel widths or for the three channel doping as in Eq. (20) (Fathil, et al. 2018):

$$m = \frac{\Delta I_D}{\Delta Q_F}$$  \hspace{1cm} (20)

3.3.1 Sensitivity of width change

In order to analyze the effect of different channel widths on the sensitivity of the biosensor, the $I_D$ versus negative $Q_F$ is plotted in Fig. 9a and b for channel width from 40 to 70 nm with p-type dopant at concentration of $1\times10^{14}$ cm$^{-3}$. With the decreasing of channel width from 70 to 40 nm, the SiNW biosensor’s sensitivity has significantly increased from 1.5 to 3.2 µA/cm$^{-2}$, and the InPNW biosensor’s sensitivity has significantly increased from 0.78 to 2.4 µA/cm$^{-2}$, respectively. These results depict more than 120% increase of sensitivity for the SiNW-FET and InPNW-FET biosensor. Be careful that the SiNW/InPNW should not be designed with very short channel width since it introduces very high resistance of SiNW/InPNW which may reduce the flow of current across the channel from drain to source.

Fig. 9 Device performance analysis $I_D$ versus negative $Q_F$ to determine the sensitivity of the a SiNW; b InPNW with different channel width of 40 nm, 60 nm, and 70 nm.
3.3.2 Sensitivity of doping change

In order to analyze the effect of doping concentration on the sensitivity of the biosensor, the ID versus negative QF for different channel doping is plotted in Fig. 10a and b. With the decreasing of channel doping from \(10 \times 10^{14}\) to \(0.1 \times 10^{14}\) \(\text{cm}^{-3}\), the SiNW biosensor’s sensitivity \((m)\) has significantly increased from 2.5 to 3.6 \(\mu\text{A/cm}^2\) and the InPNW biosensor’s sensitivity \((m)\) has significantly increased from 1.8 to 2.8 \(\mu\text{A/cm}^2\), respectively. This result depicts more than 80% increasing of sensitivity for the SiNW-FET and InPNW-FET biosensor. Be careful that the SiNW/InPNW channel should not be designed with low channel doping since it introduces very high resistance of SiNW/InPNW which may reduce the flow of current across the channel from drain to source.

Even though target analytes bind to the surface of the SiNW/InPNW, the change of \(\text{ID}\) will be kept at minimal since the SiNW/InPNW is high in resistance (Hofbauer et al. 2020). Therefore, in this simulation, the shortest SiNW/InPNW channel with the width of 40 nm and the lowest SiNW channel doping of \(0.1 \times 10^{14}\) \(\text{cm}^{-3}\) has demonstrated the best sensitivity compared to the other channels (Rani et al. 2018). The result displays good improvement in SiNW when compares to the InPNW channels, respectively. Hence, as you can see in Table 2, among the three channels with different channel width and doping which have been simulated for both of the SiNW-FET and InPNW- FET, the increasing of channel doping causes decreasing of sensitivity \((m)\) and the increasing of channel width causes decreasing of sensitivity \((m)\) and as a result the channel width of 40 nm with the lowest channel doping of \(0.1 \times 10^{14}\) \(\text{cm}^{-3}\) can be considered with high sensitivity \((m)\) for the fabrication of the actual and ideal device (Nguyen 2018).

4 Conclusion

In conclusion, the effect of different SiNW and InPNW channel width with different channel doping towards the electrical characteristics has been understood by using the device numerical modelling tool, which is Silvaco ATLAS. The decreasing of the width and doping has contributed to the increase of electrical resistance at the channel region. Due to the fact that the electrical resistance is indirectly proportional to the output current based on
the Ohm’s law, it can be concluded that with the decreasing of the channel width and doping also has caused reduction of the output $I_D$ of the NW-FET device. In addition, simulations of the biomolecules detection have been performed on the NW-FET biosensor by using negative $Q_F$ as a representation of the target analyte. The introduction of negative $Q_F$ on the surface of the channel has attracted the hole carriers inside the $p$-type SiNW, contributing to a better formation of hole conduction channel, which further increase the $I_D$ flow from drain to source region. The quantity of negative $Q_F$ determine the $I_D$ values that flows along the channel. These changes in $I_D$ is a representation of the target analyte concentration which is detected by the biosensor. The channel width of 40 nm with doping concentration of $0.1 \times 10^{14}$ cm$^{-3}$ has shown an improvement on the performance of the SiNW-FET and InPNW biosensor with the sensitivity 3.6 and 2.8 µA/cm$^{-2}$ respectively. These obtained results are for the sake of understanding the behavior of the SiNW-FET in comparison with InPNW-FET biosensor by altering its dimension and doping concentration, and may be used as a reference for the fabrication of the actual SiNW or InPNW biosensor in the future.

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