Simulation and Performance Evaluation of Charge Plasma Based Dual Pocket Biosensor using SiGe-Heterojunction TFET Design

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Abstract
Conventional biosensor designs are often vulnerable to issues like random dopant fluctuations (RDFs) and high thermal budgets due to their design and the device they are based on. The main reason behind such issues is the complexity of maintaining uniform doping levels throughout the device structure. This manuscript investigates a biosensor structure utilizing a dual pocket junctionless SiGe-Heterostructured TFET design to overcome such shortcomings. The implementation of the doping charge plasma technique with the uniform doping of $1 \times 10^{15} \text{cm}^{-3}$ along an integrated SiGe-Heterostructure layer has improved the tunneling process while also effectively eliminating issues like random dopant fluctuations (RDF). Again the overall performance also depends on the sensitivity of the sensor design. Increasing the trapping area for biomolecules at the same technological node by increasing the pocket length or by adding a pocket region leads to rapid changes in the sensor’s electric properties owing to shifting dielectric constants (k) and charge densities (in both positive and negative situations), improving in the overall detection process. The influence of these parameters on the device’s Drain current, Surface Potential, Electron Tunneling Rate (ETR), Subthreshold Swing (SS), and $I_{on}/I_{off}$ ratio is also explored. The introduction of the added pocket region gives us scalability while also showing a higher sensitivity of $5.38 \times 10^9$ for a dielectric constant being 12 and neutrally charged while rising to nearly $1.43 \times 10^{10}$ if the molecules are positively charged. With the improvement in drain current sensitivity due to the additional pocket and junctionless design, this work will undoubtedly give researchers a roadmap for the future generation of highly sensitive biosensor alternatives.

Keywords Biosensor · Charge plasma · Dielectric Modulation (DM) · Electron Tunneling Rate (ETR) · $I_{on}/I_{off}$ ratio · Sensitivity · Subthreshold Swing (SS) · Tunneling Field Effect Transistors (TFET)

1 Introduction

Humans are currently facing life-threatening alerts due to the worsened cases of biohazards in the form of HIV to present-day Coronavirus academics. Their nanoscale diameter makes them invisible, accelerating their spread at a lightning pace without knowing [1]. So proper detection of their existence and identification of biomolecules is essential to nullify such threats. The sensors have made significant progress in this area by providing a systematic method for detecting such biomolecules. They are being widely used in a wide range of industries, including medical diagnostics, medicine delivery, food processing, environmental monitoring, security, and surveillance. A biosensor can generate an electrical signal due to the internal structural change from the captured biomolecules in the pocket region [2]. The detection and the transduction are two primary phases of the sensing process of the targeted biomolecule. Analyzing targeted biomolecules in the detection stage yields electrical signals that are then processed in the transduction step to produce the final result.

The initial FET-based biosensor created a significant buzz among researchers due to its higher detection ability, lower power consumption, lower fabrication cost, and CMOS-based on-chip integration probability [3]. However, the theoretical limit of the subthreshold swing to 60mv/Dec [4], the incompetence of measuring neutral bio-molecules, and considerably greater power consumption due to the thermionic emission of carriers have led researchers to shift their focus on new FET-based biosensors [5]. Among
these, TFET stands out as an admirable device due to its lower subthreshold slope, and comparatively lower OFF current [6]. But one of the significant drawbacks of TFET design is its lower ON-current. There are several ways to address the problem: incorporating a double material gate structure, implementing heterojunction at the tunnelling interjunction to improve tunnelling probability, optimizing work function, and more [7–9]. Combining characterized lower off current with the increased ON-current, TFET shows a significant $I_{on}/I_{off}$ ratio, which improves the overall performance of the device.

The early stages of TFET-based biosensor designs used physical doping, which complicates the manufacturing process and raises the overall thermal budgeting for high-temperature thermal annealing [10, 11]. A TFET is a p-i-n structured device that works on the Band to Band tunnelling principle. As p-i-n structures are heavily doped, they are more vulnerable to issues like Random Dopant Fluctuations (RDFs) [12]. It increases the leakage current of the device due to a wide range of threshold voltage fluctuations [13]. To overcome such shortcomings, doping-less techniques are introduced [14, 15]. The doping-less approach is further subcategorized into two sections. The first focuses on the charge plasma-based method, and the other covers the electrostatic doping technique. Both of these technique requires appropriate work functions for the metal conductors to create junctions [16]. The charge plasma method involves the fulfilment of the following two prerequisites.

1. The body of a TFET device needs to maintain a body thickness less than the Debye’s length. Exceeding the parameter can cause a semiconductor band structure to respond due to the abrupt changes in the doping level. The equation for the Debye’s length can be expressed as Eq. 1 [17],

$$L_D = \sqrt{\frac{\varepsilon_S \times V_t}{q \times N_d}}$$  

(1)

Here, $\varepsilon_S$ represents the dielectric constant value for the silicon substrate, $V_t$ is the thermal voltage and the $N_d$ is the net biomolecules density present in the pocket.

2. We can choose appropriate metal electrodes with sufficient work functions for the source and drain electrodes from Eqs. 2 and 3 respectively,

$$\phi_{Source} > \left( \chi_S + \frac{E_g}{2 \times q} \right)$$  

(2)

$$\phi_{Drain} < \left( \chi_S + \frac{E_g}{2 \times q} \right)$$  

(3)

Over time the Dielectrically modulated TFET designs were also explored by the biosensor researchers to uplift the performance limit. This manuscript discusses a biosensor design that contains an additional pocket region with the conventional pocket region underneath the gate electrode to fit an additional amount of molecules at the same technological nodes. As biomolecules naturally have different dielectric constants from the oxide layer of the sensor and so, the presence of the biomolecules at an efficient amount alters the electrical characteristics, which can further identify the molecules. In contrast with the sensor design, the TFET upon which the device is based also needs to have performance aspects of supporting the design improvements. As TFETs are known to have lower ON currents, and the device needs to have a near intrinsic doping level, a SiGe-Heterostructure approach is taken to improve its performance. Choosing a lower bandgap at the source channel junction enables a higher number of carriers to tunnel through compared to homojunction, improving the current. So, in summary, the work focuses on a SiGe-heterojunction TFET-based dual pocket Biosensor design that helps the tunnelling process and increases the sensor’s sensitivity by accommodating significantly higher Biomolecules for better performance.

The basic structure and design are covered in Section 2. In Section 3, the effect of the variety of biomolecules present in the pocket area, the charge density of the molecules, and the device shape on the performance of the device, fill factors and uniform or non-uniform accommodation of biomolecules is explored in depth. In Section 4, a sensitivity analysis with respect to multiple parameters is presented, and finally, in Section 5, the conclusion is made.

## 2 Device Structure and Simulation Framework

The structural approach of the proposed Dual pocket Charged Plasma Based Biosensor using SiGe-Heterojunction is shown in Fig. 1. All necessary parameters regarding its design architecture are presented in Table 1. Here we have gone for a symmetrical length for the source, drain, and the channel length at 80 nm. A metal electrode with a higher bandgap is recommended to create a strong P+ region at the source region. As such, a platinum electrode with 5.93 eV work function is utilized as the source electrode. Similarly, to introduce a strong N+ region at the drain side, a metal electrode (hafnium) with a work function of 3.8 eV is utilized. The doping-less approach also simplifies the manufacturing process and reduces the overall thermal
budget. Our proposed device’s Silicon body thickness ($T_{Si}$) is kept at 10nm. The choice for this parameter is made keeping the Debyes length in mind, which is more prominent in every experimental situation. A gap of 3.5nm is kept for the biomolecules to get in and be trapped in the pocket region between the gate and source. Such air gap was chosen to be as small as possible between the source and the gate to aid the tunnelling process. Between the source material (SiGe) and the source electrode, an extremely thin layer of SiO$_2$ is maintained [18]. This attempt prevents generating Silicade particles when the source electrode is in direct contact with the SiGe source [19]. Two separate metal electrodes are used for the gate electrode containing different work functions. The gate electrode near the source region is Magnesium having a work function of 3.7eV, and the electrode facing the drain region is chosen to be gold with a work function of 5.1eV. This approach guarantees the lower OFF state current since $I_{off}$ is dictated by thermionic emission instead of Band to Band Tunneling (BTBT). Again a portion of the HfO$_2$ layer is used as a high dielectric material next to the pocket area to increase the drain current value due to its more significant dielectric constant.

We have used the Silvaco Atlas platform for the simulation stage for the performance analysis of the biosensor design. A variety of simulation models and approaches were considered to obtain desired outcomes. In the end, the drift-diffusion transportation model combined with the Shockley-Read-Hall (SRH) Recombination was utilized. The Bandgap narrowing model (BGN) is used to capture the effects of Bandgap Narrowing. This is because it is seen that in semiconductor materials, the separation between the conduction band and the valence band starts to decrease from the default characterized value due to the effects of the higher doping levels, typically starting from $10^{19}$cm$^{-3}$. The Lombardi CVT and the Non-local Band to Band Tunneling model are utilized to capture the impact of the tunneling accurately. Again some other parameters regarding the simulation process and the materials used during the simulation period, like the Lifetime of minority and majority carriers, electron and hole densities, etc., are turned into the parameter file for optimum tool calibration. Figure 2 shows the effects of the transfer characteristics and the sensitivity curve of the dual pocket design against a single pocket variation of the biosensor while all described models are being utilized.

![Fig. 1 Structure of proposed Dual-Pocket Charge Plasma Based Biosensor using SiGe-Heterojunction TFET](https://example.com/f1.png)

**Table 1 Specifications for Dual Pocket Charge Plasma Based Biosensor using SiGe-Heterojunction TFET**

| Parameters                              | Parameter Symbol | Values specified          |
|-----------------------------------------|------------------|---------------------------|
| Doping at Channel Region                | $N_c$            | $1 \times 10^{15}$ atoms/cm$^3$ |
| Oxide Thickness                         | $T_{ox}$         | 5nm                       |
| Silicon Body Thickness                  | $T_{Si}$         | 10 nm                     |
| Work Function of the source electrode   | $\phi_{M1}$     | 5.1eV                     |
| Work Function of the drain electrode    | $\phi_{M2}$     | 3.8eV                     |
| Thickness of Pocket/Cavity region       | $T_{pocket}$     | 5nm                       |
| Length of Pocket/Cavity region per unit | $L_{pocket}$     | 35nm                      |
| Air space for Pocket/Cavity             | $L_{air}$        | 3.5nm                     |
| Thickness of HfO$_2$ dielectric region  | $T_{HfO_2}$      | 5nm                       |
| Length of HfO$_2$ dielectric region     | $L_{HfO_2}$      | 45nm                      |
3 Results and Discussions

The captured biomolecules at the pocket area usually have various dielectric constants ranging from 1.54, indicating uricase, to 2.63 as biotin and much more [20–23]. The captured biomolecules are neutral by default, but they are not uncommon to be positively or negatively charged. The effects of various bio-molecules parameters such as dielectric constant and variation of the charge density over the tunneling process of the Fet are investigated in this section. As the tunneling rate of the device increases, more electrons tend to pass the source to the channel tunneling barrier, increasing the total current. With the Wentzel-Kramer-Brillouin (WKB) approximation, given by Eq. 4, the tunneling rate effects may be understood in a more clear and concise manner [24].

\[ P(T) = \exp \left( \frac{-4\lambda \sqrt{2m^*E_g^{1.5}}}{3|e|h(E_g + \Delta \phi)} \right) \Delta \phi \]

(4)

Here \( m^* \) and \( E_g \) represent the effective mass and bandgap of the material, while \( e \) and \( h \) denote the charge of electrons and the plank’s constant. \( \Delta \phi \) represents the energy difference between the conduction band of the source side and the channel side’s valance band. \( \lambda \) represents the value of the tunneling length of the electron, which can be expressed as,

\[ \lambda = \sqrt{\frac{\varepsilon_{Si}}{\varepsilon_{ox}t_{Si}t_{ox}}} \]

(5)

Again \( \varepsilon_{Si} \) and \( \varepsilon_{ox} \) depict the relative permittivity value of the oxide layer and the silicon layer accordingly. While \( t_{Si} \) and \( t_{ox} \) represent the thickness of the oxide layer and the channel section.

From Eq. 4, it is prominent that to improve a device’s tunneling probability, the tunneling length(\( \lambda \)) has to be reduced as they contain an inverse relationship. The tunneling length can be shrunk down by lowering the thickness of the oxide layer, otherwise raising the dielectric constant value. However, reducing the oxide layer can introduce a more significant leakage current at the gate. So the alternative is preferable. Improved drain current for different biomolecules can significantly improve the device’s sensitivity, hence improving the sensor performance.

3.1 Effects of Dielectric Constant Variation

The dielectric constant of the biomolecules trapped in the pocket area is the primary focus of the initial investigation of the device’s performance. Due to their dielectric values, their presence alters the electrical properties of the proposed biosensor sensor, which can be observed through the transfer and output characteristics represented in Fig. 3. The dielectric constants of a few practically available biomolecules are presented in Table 2. However, for the simplicity of the simulation process, a range of dielectric constants starting from 1 to 12 is considered. Here \( k=1 \) indicates the pocket region to be filled with air. It is used as a baseline to measure the sensitivity of the other dielectric constants. Again \( k=2 \) indicates biomolecules present in the pocket region containing a dielectric constant of 2 and so on. In this stage, the charge density for each situation is considered a neutral stage. As per the roadmap demonstrated by the IRDS, the scaling of the Effective Oxide Thickness (EOT) can maintain electrostatic control over the channel area. Dielectric materials with higher dielectric value can improve
the electrostatic control without reducing the oxide thickness. Equation 6 reveals that effective higher gate capacitance can improve the coupling capacitance between the gate and the tunneling junction. It provides a strong ON-current, making it feasible to detect particular interactions between biological molecules [25]. Improved control over the channel region improves the overall performance of the device. Here, the ON current for k=12 is around $3.10 \times 10^{-6}$ A/μm, and for the empty pocket region, the ON current comes down to $6.01 \times 10^{-13}$ A/μm. Such a significant differentiation can easily help accurately differentiate the molecules, improving the liability of the sensor.

Figure 4(a) holds the effects of the dielectric modulations over the energy band of the device. In every situation, the $V_{gs}$ was set at 1.5V, and the $V_{ds}$ was set at 0.7V. According to the graph, the dielectric value increases the band bending in the source channel junction more prominent. A less comprehensive source to channel region helps considerably more carriers tunnel through, increasing the ON current. A similar effect can also be observed in Fig. 3(a): the biosensor’s transfer characteristics curve with varying dielectric constants and neutral charge density in the pocket region. Here Eq 7 can formulate the Band to Band tunneling rate of carriers [18].

$$G_{BTBT} = A \frac{E^2}{E^2 + E^2} \exp \left( - \frac{BE_k}{E} \right)$$

The bandgap for the semiconductor material is represented by $E_g$ in this equation, while $E$ stands for the electric field. Again, both A and B are material-based factors. As the Electric field increases, the $G_{BTBT}$ rate will increase, resulting in a considerable ON current improvement.

Figure 5 demonstrates the effect that altering the dielectric constant (k) will have on Electron Tunneling Rate (ETR), Subthreshold Swing (SS), and $I_{ON}/I_{OFF}$ ratio. When the constant value (k) is changed between 1 and 12, SS will drop, resulting in a shorter reaction time. Stronger interaction between the gate and channel regions is the cause. However, ETR and $I_{ON}/I_{OFF}$ ratios improve when the tunneling barrier decreases due to a rise in dielectric constant value. Table 3 also shows the statistics of SS, $I_{ON}/I_{OFF}$ ratio values, and ETR under the effects of dielectric modulation.

![Graph](image_url)
3.2 Effects of Charge Density Variation

The biomolecules present in the pocket region being positively or negatively charged can alter the sensor's electrical properties, which also need to be considered.

The gate to source voltage for a FET structure under the influence of charged molecules can be derived as Eq. 8.

\[
V_{GS} = \Psi_S + \Phi_{MS} - \left( \frac{q(\pm N_{bio})}{C_{eff}} \right)
\] (8)

Fig. 4 Impact of varying dielectric constants on (a) Energy Band diagram and (b) Surface potential with neutral biomolecules present in the pocket region.

Fig. 5 Impact of varying dielectric constants on (a) Electron Tunneling Rate (ETR) and Subthreshold Swing (SS) (b) $I_{ON}/I_{OFF}$ ratio with neutral biomolecules present in the pocket region.
Here, $\Psi_S$ holds the surface potential value, while $\Phi_{MS}$ represents the difference between metal and semiconductor work functions. The $N_{bio}$ represents the value of the charge density of the biomolecules within the pocket area. When the charge density is positive, then the equation converts into,

$$V_{GS} = \Psi_S + \Phi_{MS} - \left( \frac{qN_{bio}}{C_{eff}} \right)$$

Equation 9 shows if the $V_{GS}$ is kept at a fixed state and the value of the positive charge density is increased, the third segment $\left( \frac{qN_{bio}}{C_{eff}} \right)$ of the equation tends to increase.

To neutralize the situation, Surface potential($\Psi_S$) has to increase. The increased surface potential boosts the tunnelling rate (ETR), which improves the drain current. Figure 6 (a) demonstrates the impact. The transfer properties improve as the positive density of the Biomolecules present in the pocket gradually increases. Again, if negative charge density biomolecules are present in the nano-cavity area, Eq. 8 transforms to,

$$V_{GS} = \Psi_S + \Phi_{MS} + \left( \frac{qN_{bio}}{C_{eff}} \right)$$

(10)

So as per the previous situation, gradually increasing the negative charge density will ultimately reduce the third term $\left( \frac{qN_{bio}}{C_{eff}} \right)$ of the equation. To balance it out, the surface potential ($\Psi_S$) is also reduced, ultimately limiting the tunnelling process. Lower tunnelling rate of the device results in a lower drain current which can also be seen in Fig. 6(b).

Figure 7 shows the impact of varying the charge density of the biomolecules present in the pocket region over the $I_{ON}/I_{OFF}$ ratio and Electron Tunneling Rate (ETR). When the density of the biomolecules is positive and raised from 0 to $10^{12}$, the ETR shows a linear increment, increasing the $I_{ON}/I_{OFF}$ ratio. In contrast, a similar linear decrease rate is observed when the density of the negatively charged biomolecules is increased. The decreased ETR also lowers the $I_{ON}/I_{OFF}$ ratio.

### 3.3 Effects of Geometry Parameters Variation

This subsection represents the impact of varying geometrical parameters of the biosensor device like the thickness and pocket length through the transfer characteristics curve in Fig. 8. The length of the pocket is kept constant at 35 nm while the thickness of the pocket area is changed from 3 nm to 6 nm in Fig. 8(a). When the pocket area becomes greater,
the barrier that the carriers have to overcome in order to tunnel through gets tougher. As a result, fewer carriers can pass through, which affects the ON current, ultimately decreasing the sensor’s overall sensitivity. Again, when the pocket region’s length is increased or decreased, keeping the thickness at a constant value, due to the Band to Band Tunneling (BTBT) mechanism of current conduction of a Junctionless TFET device, there will be marginally no change in the Transfer characteristics. Figure 8(b) illustrates the effect when the device’s length is inspected across the range of 25nm to 55nm as all of them are overlapping each other.

Again when the biomolecules present in the pocket region are changed from the dielectric constant of $k=1$ to 12 at a constant gate voltage of 0.7V, the results can be seen in Fig. 9. As the pocket thickness is lower at 3nm due to its higher $I_{ON}/I_{OFF}$ ratio observed at the previous stage, the curve for this particular point shows competitively higher values than others. However, as the thickness increases, more biomolecules can be facilitated in the pocket region, ....
increasing the separation between \( k=1 \) and \( k=12 \) values. This increase in separation results in better sensitivity, expressed in detail in a later section of the manuscript. Again a similar explanation can also be given in the case of the tunnelling rate. With the ease of carrier tunnelling at a thickness of 3nm, the values are pretty higher; however, the greater separation between each dielectric change is visible as more biomolecules can enter the pocket region.

### 3.4 Effects of Fill Factor Variation

The fill factor is an essential factor for determining the performance aspect of a Biomolecule design. But often, the pocket region can not be completely filled up due to the steric hindrance. The situation can affect the performance of the device. Up to this point of the study, the pocket region has been considered filled up, indicating a fill factor of 100%. Anything less than that has been considered partially filled. Again, the possibility of both uniform and non-uniform fill of biomolecules must also be considered in a partially filled pocket region. Here, Fig. 10, shows a visual representation of the difference between the fully filled pocket region with the partially and uniformly filled pocket region. Here, in the fully filled pocket, the length of the pocket region and the length accommodated by the biomolecules are to be the same as \( L_{\text{pocket}} = L_{\text{bio}} \). In contrast, in the case of the uniform partial case, the length of the pocket region is larger than the length accommodated by the biomolecules resulting in \( L_{\text{pocket}} > L_{\text{bio}} \). In this situation, the rest of the place is filled with air.

Again the effective capacitance of the device also starts to change from the pocket region being fully filled to partially filled situations. In Fig. 11, we can see a representation of the effective capacitance of fully and partially filled cases.

1. If we consider the pocket region to be completely filled with desired biomolecules, then the effective capacitance per unit area \( (C_{\text{eff}}) \) of the pocket region is,
1. Here, $\varepsilon_{\text{SiO}_2}$ and $\varepsilon_{\text{bio}}$ represent the dielectric constant value of the thin silicon dioxide layer under the pocket region and the value of the biomolecules present in the region, respectively. $t_{\text{SiO}_2}$ illustrates the thickness of the thin oxide layer.

2. If the pocket region is partially and uniformly filled, there will be two regions in the same pocket area. One partition of the region will be filled with biomolecules, and in the other, there will be air. In this case, the effective capacitance will also be divided into two parts. The capacitance for the biomolecule portion is,

$$C_{\text{eff}} = \frac{\varepsilon_{\text{SiO}_2} \times \varepsilon_{\text{bio}}}{(\varepsilon_{\text{SiO}_2} \times \varepsilon_{\text{bio}}) + (\varepsilon_{\text{SiO}_2} \times t_{\text{SiO}_2})}$$

(11)

Here, $\varepsilon_{\text{SiO}_2}$ and $\varepsilon_{\text{bio}}$ represent the dielectric constant value of the thin silicon dioxide layer under the pocket region and the value of the biomolecules present in the region, respectively. $t_{\text{SiO}_2}$ illustrates the thickness of the thin oxide layer.

Now to analyze the behavior of a partially filled pocket in a uniform formation, five different fill factors (100%, 70%, 50%, 20%, 10%) are considered for judging overall performance. The transfer characteristics of the observation are illustrated in Fig. 11(a). Here we can see that the maximum drain current is observed in the case of a 100% fill factor. This is understandable as more amount of biomolecules will have a strong influence on the electrical changes that appear in the device. Decreasing amount of biomolecules present in the pocket shows the downward trend of the drain current. The lowest amount of drain current is seen in the case of the 10% Fill factor, which is the lowest amount considered for this study. Figure 11(b) also shows a similar representation of the previous observation in terms of sub-threshold swing and $I_{\text{on}}/I_{\text{off}}$ ratio. At the 100% both subthreshold swing and the $I_{\text{on}}/I_{\text{off}}$ ratio better value in comparison to the lowest fill factor.

Again, the position of the biomolecules can also cause performance alteration. The drain current is more significant if the biomolecules are accumulated near the gate region rather than opposite the gate. Figure 12 shows the effect of this cause. When the pocket region is partially filled (In this case 50% and 20%), if the molecules are accumulated near the gate region, it has a more significant influence on the tunneling process in the source to channel tunneling process instead of the biomolecules being in the opposite direction. As a result, the drain current is seen to be greater than the opposite case.

Again, Fig. 13(a) shows the energy band diagram for a fill factor of 50% with two different position possibilities. The tunneling barrier is lesser when the biomolecules...
are accumulated near the gate. This is due to the more significant influence of the biomolecules remaining near the gate. As a result, in the opposite case, the tunneling barrier is much broader. This figure also supports the increased drain current in case the biomolecules gather near the gate as more carriers can tunnel through. Again due to more significant influence, the surface potential is sharper than the alternating case resulting in a better current rate.

Although uniform distribution has already been discussed earlier in this section, biomolecules can also infiltrate the pocket region in a non-uniform manner, and also this is the most likely scenario created in real life. So the performance aspect of such a situation must also be analyzed. In this study, we have considered four individual non-uniform profiles, which are a) Increasing, b) Decreasing, c) Concave, and d) Convex. Here the Fig. 14, shows a visual presentation of the individual profile inside the pocket region. Herein each case, the Fill Factor is kept at 60% to have an equal floor to analyze their performance aspects.

As described before, when the biomolecules are accumulated near the junction region, it significantly influences the tunneling process, letting a higher number of carriers tunnel through. The same effect can also be observed in the non-uniform profile. From Fig. 14, we see that the increasing and the Concave profile has the most amount of biomolecules near the junction, resulting in a more significant drain current. This effect can be seen in Fig. 15(a), where the transfer characteristics are shown for the different non-uniform profiles. The decreasing and the Convex are falling behind in the drain current amount due to the biomolecules accumulating further from the junction. The effects can also be observed in Fig. 15(b), where the Subthreshold swing and the $I_{on}/I_{off}$ ratio are shown. As described, the Increasing and Concave perform better than the decreasing and Convex profiles.

4 Sensitivity Analysis

Sensitivity is one of the detecting factors in determining the performance of a Biosensor device. Biomolecules are often relatively tiny in size. In often cases, small fragments
of DNA-sized biomolecules are intended to be measured in the size of 3 to 4nm. Successful detection of such a small-sized molecule’s presence is achievable only if the sensor’s sensitivity is high enough [26]. When the molecules enter the pocket region, it alters the electrical properties of the TFET device, which helps the detection process.

\[
S_{I_D} = \left( \frac{I_{D_{bio}} - I_{D_{air}}}{I_{D_{air}}} \right) \tag{15}
\]

Here Eq. 15 stands for the basic operation of the sensitivity measurement of a Biosensor device. The \(I_{D_{bio}}\) means the drain current of the sensor achieved through biomolecules...
being trapped in the dual pocket area, while \( I_{D_{\text{air}}} \) is the drain current amount when the dual pocket is empty of molecules.

In contrast to the sensitivity derived by the drain current, the Subthreshold sensitivity can also be considered a significant parameter in the performance judgment of a biosensor design. The parameter is used to successfully demonstrate a sensor’s detection speed, which is also essential.

\[
S_{SS} = \left( \frac{SS_{\text{bio}} - SS_{\text{air}}}{SS_{\text{air}}} \right)
\]  

Equation 16 represents a mathematical representation of the Subthreshold Swing sensitivity \((SS_{SS})\) where the notation \(SS_{\text{bio}}\) depicts the subthreshold swing of the biosensor when the pocket region is populated with neutral or charged (positive or negative) biomolecules. Again, \(SS_{\text{air}}\) denotes the situation when the pocket of the sensor is considered empty.

Figure 16 depicts the improvements in sensitivity with the increasing value of dielectric constants. Increasing the constant value also increases the electrostatic control over the gate without physically reducing the oxide thickness, improving the transfer characteristics curve for a constant drain to source voltage. An improved transfer characteristic helps to enhance the sensitivity of the particular molecule against air, ultimately improving the sensor performance. Here The sensitivity curve for the dielectric constants 2, 6, 7, 10, and 12 is constructed since the model performs precisely for the lower dielectric constant values.

Again if the charge density is shifted from neutral to positive or negative, the sensitivity is also affected. The sensitivity of the sensor is shown to have improved when a positive charge density is applied from \(1 \times 10^{11}\) to \(1 \times 10^{12}\). However, an opposite effect is observed when the negative charge density is considered for the biomolecules. The sensitivity decreases with the increase of negatively charged biomolecules. These effects are shown in Fig. 17.

Figure 18 represents the effects of Subthreshold Swing Sensitivity on positive, neutral, and negatively charged biomolecules, respectively, with the variation of the dielectric constant. As the value increases from 2 to 12, the columns of the \(S_{SS}\) also increasing constantly. A higher number value indicates a superior detection probability and
increased electrical response to the particular biomolecules in the pocket region. This observation can also be verified in Fig. 10, showing the drain current sensitivity. As the dielectric value increases, the value of the sensitivity increases, which also means the sensor is more likely to detect the molecule more precisely hence the increased SSS for the constant. Again the changing density also has there own set of effects over the SSS. The biomolecules present in the pocket region being charged also show a slight deviation in values from naturally charged scenarios. Taking (k=12) under consideration, the value of the SSS has shown a value of less than 0.2 for neutrally charged scenarios, while in the case of positively charged, the value levels up to greater than 0.2, indicating better detection probability. Again, if the molecules are negatively charged, the value of SSS falls below 0.15. The observation also supports the graphs in Fig. 17 from earlier.

However, a situation involving both variables is created to better understand the effects of different charge densities with various dielectric constants. The effects of dielectric constant variation, including the charge density of the biomolecules included in the pocket region over the device’s sensitivity, are observed in Fig. 19. Here, situations with the pocket region filled with positively charged biomolecules show comparatively higher sensitivity than neutral situations. In contrast, cases with negatively charged biomolecules show relatively lower sensitivity. The estimated sensitivity of prior published experiments on bio-sensors is compared in Table 4. The Proposed design of the dual pocket charge plasma-based Si-Ge heterojunction biosensor has a maximum sensitivity of \(5.38 \times 10^9\) for the dielectric constant of 12 at neutral density.

But if the positive charge density is considered, the sensitivity rises up to \(1.43 \times 10^{10}\).

The fill factor has also shown its effects on the Sensitivity value of the device, which can be observed in Fig. 20. The fill factor indicates the percentage of the pocket region filled with biomolecules. So a higher rate will undoubtedly hold a greater sensitivity. Figure 20(a) shows a gradual increase in the sensitivity value as the total percentage of the biomolecule in the pocket region goes up towards 100%. Again, the biomolecules also can be in a non-uniform manner in a partially filled situation, which can affect the performance aspect of the device. Here we have considered four non-uniform profiles, and their sensitivity can be seen in Fig. 20(b). The Increasing and concave profile has greater sensitivity due to the closer position of the biomolecules towards the junction of the device, assisting the tunnelling process. And due to the opposite reseason, the decreasing and the Concave profile shows less sensitivity.

| Publication Year | Study Reference                | Approximate Sensitivity |
|------------------|--------------------------------|------------------------|
| 2022             | This Work                      | \(1.43 \times 10^{10}\) |
| 2021             | Dewan et al. [27]              | \(1.13 \times 10^{10}\) |
| 2021             | Vimala1 et al. [28]            | \(2 \times 10^8\)      |
| 2021             | Chong et al. [29]              | \(1.38 \times 10^5\)   |
| 2017             | Noor et al. [30]               | \(4.82 \times 10^7\)   |
| 2016             | Sing et al. [31]               | \(9.48 \times 10^5\)   |
| 2015             | Narang et al. [26]             | \(3 \times 10^6\)      |
| 2015             | kanungo et al. [32]            | \(1 \times 10^6\)      |
5 Conclusion

This work represents a performance investigation of the proposed charge plasma-based dual pocket biosensor using Si-Ge heterojunction junctionless TFET. The random dopant fluctuations (RDFs) phenomena and difficulties with high thermal budgeting were prevented because of the approach of charged plasma technology. Again, using a lower bandgap material like the SiGe at the source end helps the tunneling rate from the source to the channel junction. The impact that changing the dielectric constant has on drain current, Electron Tunneling Rate (ETR), Subthreshold Swing, $I_{on}/I_{off}$ ratio, and Electric field, is also investigated in great detail. The results show that raising the dielectric constant of the biomolecules in the pocket area improves the device’s ON-state current while keeping the OFF-state current at a constant state. Due to this, an $I_{on}/I_{off}$ ratio of $2.87 \times 10^{11}$ has been achieved with a dielectric constant of $k=12$ being present in the pocket area. Positive, negative, and neutrally charged biomolecules present in the pocket region were also simulated, and their effects on device performance. The on-state current of the device increases (decreases) when the pocket is populated by positively (negatively) charged biomolecules, according to the results. The effects of different pocket geometry were also investigated over the sensor performance. In the end, during the sensitivity analysis, it is observed that when neutrally charged biomolecules occupy the nanocavity area with a dielectric constant of 12, then a maximum sensitivity of $5.38 \times 10^9$ is reached. Again if the pocket region is occupied with positively charged biomolecules, sensitivity can rise to $1.43 \times 10^{10}$. Finally, the suggested study is compared to previously published studies regarding their sensitivity.

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Declarations All procedures conducted in research involving human participants complied with ethical requirements.

Ethical Approval Approval is assured from the author.

Consent to Participate Consent is assured from the author of the study.

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