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Design of Field Effect Transistor Biosensor based on Graphene Nanoribbons with High Resolution∗

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Abstract

Graphene with ultrasensitive, real-time and label-free electrical detection has attracted interests to be used as a bioelectronic platform for sensing biomolecules in different fields. Using graphene nanoribbons solves zero gap energy problem of graphene and makes it suitable to use directly in transistor applications such as field effect transistors. Researchers are always looking for the high sensitivity of biosensors to detect low concentrations of biomolecules in biological samples. In this work, by introducing two important parameters: width of ribbons and portions of first three subband indexes in conductance, an analytical model is proposed for graphene nanoribbon’s conductance. Here for the first time, the sensitivity of transistors to detect low concentration of biomolecules as a powerful and flexible electronic biosensor is controlled by using the width of graphene nanoribbons. Finally, FET device as a biosensor according to an experimental work to detect concentration of biomolecules, charge and thickness of membranes is calibrated.

Keywords: Graphene Nanoribbons, Field Effect Transistors, Biosensors, Conductance.

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1. Introduction

Graphene, a single layer of graphite, in the last decade has been discovered as an attractive material with low dimensional physics for applications of carbon-based electronics and biosensing devices [1, 2, 3, 4, 5, 6]. Hybridized $sp^2$ bonding makes graphene structure look like a honeycomb crystal. So, the first Brillouin zone is hexagonal with six $K$ points corresponding to six corners of the Brillouin zone that only two of these points are inequivalent. These two distinct $K$ points, commonly labeled $K$ and $K'$, called Dirac points where it is because of their unique position in the linear energy spectrum of graphene.

Two dimensional (2D) nature of graphene leads to a unique electronic structure as a linear dispersion relation at $K$ points in the Brillouin zone. In this structure, the charge carriers known as massless Dirac fermions (zero effective mass) lead to interesting electronic properties such as room-temperature am-bipolar characteristics, i.e., the charge carriers can be alternated between holes and electrons depending upon the applied gate voltage [7, 8]. As graphene is characterized as a semi-metal or zero gap semiconductor, it is not suitable to use directly in transistor applications such as field effect transistors (FET). However, opening a band gap in this material gives rise to novel methods [9]. The energy gap can be induced by confining the electrons in graphene. Graphene nanoribbons (GNRs) are a novel and intriguing class of materials in the field of nanoelectronics [10, 11]. When graphene is patterned into a narrow ribbon with nanometer size, the charge carriers are confined to a quasi-one-dimensional system, thus the energy gap could be created [12, 13]. Depending on the boundary conditions, some sets of 1D modes do not pass through the intersection point of the conduction and valence band, so these quasi-1D graphene ribbons become semiconductors with a finite energy gap as a function of their widths ($W$) [14, 15]. GNRs with two shaped edges, zigzag and armchair configuration, show different electrical properties. Zigzag GNRs have metallic structures, while the GNRs with armchair shaped edges can be either metallic or semiconductor depending on their widths. The armchair GNRs are classified into three different
families: \( N_a = 3m + 2 \), \( N_a = 3m + 1 \) and \( N_a = 3m \), where \( N_a \) is the number of carbon atoms across the ribbon width and \( m \) is an integer \((m = 0, 1, 2, 3, \ldots)\) \[15\]. Here, \( N_a = 3m + 2 \) family is metallic as zero band gap, while \( N_a = 3m + 1 \) and \( N_a = 3m \) family are semiconductors which their band gap can be scaled inversely with the width of ribbons, as report in some experimental works \[16,17\].

As the performance of FET biosensors depends on properties of the channel material and the biosensor structure, GNR with high sensitivity to electronic perturbations from foreign biomolecules makes it an ideal material to use directly in FET channel biosensors. Also, GNR coated with lipid membranes or composite polymers are suitable for using in bioapplications as hybrid devices \[18\].

When GNR doped with charge impurities, its conductance is influenced by the decreased or increased density of carriers, that comparing the change in position of the Fermi energy (the Dirac point shifting) as a signal to detect charged biomolecules.

There are lots of theoretical and experimental work based on the Dirac point shift to detect various target molecules in biological samples. Dong \textit{et al.} \[19\], reported a sensitive graphene-based FET for label-free electrical detection of DNA by evaluation of the Dirac point shift. Graphene FETs as devices with high sensitivity to pH changes in aqueous solution have been already studied by Kim \textit{et al.} \[20\]. They used graphene-Ag nanowire hybrid in FET to detect protein. Also, the graphene-FET biosensors for sensing glucose concentrations have been studied by Kwak \textit{et al.} \[21\], to measure drain-source current and the Dirac point shift.

However, it should be noted that intrinsic graphene as a zero band gap material exhibits a low on/off current ratio when used as a FET channel, that limits the sensitivity in biosensors. Tan \textit{et al.} \[22\] investigated the edge effects of GNR on responses to different PH in electrolyte gate FET by evaluating the Dirac point shift. Tamersit \textit{et al.} \[23\] showed that FETs based on a single GNR have a high sensitivity and resolution in biosensors. Takashima \textit{et al.} \[24\] used the nonequilibrium Green’s function method to investigate the effects of edge disorder of GNR on FET characteristics.
New methods and models are still needed to study how one can improve sensitivity and resolution of devices when they function as biosensors. On the other hand, so far no effort has been made to provide a comprehensive understanding the effects of different GNR widths on biosensors sensitivity through a concept of Dirac point shift. To the best of our knowledge, this work is the first study, to model the conductance of un-doped and doped GNRs as a function of energy gap and Fermi energy, to detect low concentrations of biomolecules with high resolution.

The main goal of this work is to express the effects of the GNR width as a parameter to control biosensors sensitivity to detect low concentrations of biomolecules with high resolution. For this purpose, this paper is organized as follows: First, an analytical conductance model for an un-doped GNR, as a function of the Fermi energy and energy gap, is introduced. Then, the portion of first, second and third conduction or valence band of the GNR is shown on conductance. And also, the effects of different GNR widths on resolution in FET to control the device sensitivity are investigated. Here, we choose GNR with widths ranging from 4 - 40nm according to reports in some experimental studies. Finally, we focus to modify conductance model for doped GNRs channel in FET, while the device is calibrated as a biosensor to detect concentration, charge and thickness of the membranes.

2. Analytical Model

The band energy of graphene using the tight binding approach that considers only the first nearest neighbor interaction, is

\[ E(\vec{k}) = \pm t \sqrt{1 + 4\cos \left( \frac{3k_xa_c}{2} \right) \cos \left( \frac{\sqrt{3}k_ya_c}{2} \right) + 4\cos^2 \left( \frac{\sqrt{3}k_ya_c}{2} \right) + 4\cos^2 \left( \frac{\sqrt{3}k_ya_c}{2} \right)}, \quad (1) \]

where \( t = 2.7eV \) is the nearest neighbor hopping parameter, \( a_{c-c} = 0.142nm \) is lattice constant and \( k_x \) and \( k_y \) are wave vector components in the \( x \) and \( y \) direction, respectively. When the electrons in \( 2D \) graphene are restricted to a
quasi-1D GNRs, the $k_y$ will be discretized along the width of GNRs. In low energy limit for the graphene band structure due to the approximation near the Fermi point [25], the Eq. (1) re wrote as like

$$E \approx \pm \frac{3}{2} t_{ac-c} \sqrt{k_x^2 + \xi_y^2};$$  \hspace{1cm} (2)

$$\xi_y = \frac{2\pi}{\sqrt{3}a_{c-c}} \frac{P_i}{(N_a + 1)} - \frac{2}{3},$$  \hspace{1cm} (3)

here ± denote the conduction and valence band respectively, $\xi_y$ is discretized wave vector in the $y$ direction, where $P_i$ stands for subbands in either valence or conduction band, and $N_a$ is number of dimer line along the width of ribbon. Fig. 1 shows the schematic structure of the armchair and zigzag GNR respectively.

We can define the width, $W$, for armchair GNRs by the number of carbon atoms along the GNR width as following

$$W = (N_a - 1) \frac{\sqrt{3}}{2} a_{c-c}.$$  \hspace{1cm} (4)

Using the Landauer formula [28], the conductance for devices based on graphene can be expressed by

$$G = \frac{2q^2}{h} \int_{-\infty}^{\infty} T(E) \frac{M(E)}{KT} \left(-\frac{\partial f(E)}{\partial E}\right) dE,$$  \hspace{1cm} (5)

where $q$ is electron charge, $h$ is Plank constant, $M(E)$ is number of modes along the length of the GNR in $x$ direction, $f(E)$ is Fermi-Dirac distribution function, $K$ is Boltzmann constant, $T$ is temperature in Kelvin, and $T(E)$ is transmission probability of the charge carriers in each subband where it is unity in ballistic limit for graphene. Room-temperature ambipolar characteristics, i.e., the charge carriers can be alternated between holes and electrons depending upon the nature of the gate voltage. So we can split not change conductance formula into two parts; one corresponds to valence band ($G_V$), and another is referred to conductance band ($G_C$). Now, we can calculate the number of modes as
where \( L \) is length of nanoribbon in \( x \) direction. So we can write Eq. 6 for valence \( (M_V(E)) \) and conduction bands \( (M_C(E)) \), respectively, as

\[
M_V(E) = \left( -\frac{3ta_{e-c}}{2L} \right) \sqrt{E^2 - \left( \frac{E_g}{2} \right)^2},
\]

\[
M_C(E) = \left( \frac{+3ta_{e-c}}{2L} \right) \sqrt{E^2 - \left( \frac{E_g}{2} \right)^2},
\]

here, we assume that \( E_g = 3a_{e-c}e\xi_y \) is energy gap.

Now, by replacing Eq. 7 and Eq. 8 into the Eq. 5, we have the total conductance model for GNR as

\[
G = \frac{2\pi^2 (3ta_{e-c})}{h(2LkT)} \left( \int_{-\infty}^{EV} \frac{\sqrt{X^2 - g^2}}{X} \left( \frac{e^{f-X}}{1 + e^{f-X}} \right)^2 dX + \int_{EC}^{\infty} \frac{\sqrt{X^2 - g^2}}{X} \left( \frac{e^{X-f}}{1 + e^{X-f}} \right)^2 dX \right),
\]

where we used \( X = \frac{E}{kT}, \ g = \frac{E_g}{2kT} \) and \( f = \frac{E_f}{kT} \) that \( E_f \) is Fermi energy. This model introduces the conductance of un-doped GNRs as a function of Fermi energy \( (E_f) \) and energy gap \( (E_g) \). Here, band gap inversely is proportional to the width of GNR, and Fermi energy as a function of gate voltage plays an important role in GNR conductance. By introducing electrical charge on GNR (doped GNR), Fermi energy will be modified, and as a result, a voltage will be
induced in the system.

According to Eq. 9, the conductance of a GNR with 49 nm width and 1 µm length, for $0.1v < V_g < 1v$ varies as $2\mu s < G < 18\mu s$ for an un-doped GNR at $T = 200K$. In such a case, the energy gap is about 23 meV, while by decreasing the ribbon width a large gap region appears near the charge neutrality point (Dirac point) where the carriers are confined to a quasi-1D system. These results are in agreement with report of Han et al. [12] that they investigated the effects of GNR width and temperature in conductance and energy gap.

3. Results and Discussion

3.1. GNR based on Field Effect Transistor

In the framework of presented model, the conductance for three families of armchair GNR with respect to the first, second and third conduction or valence band are investigated versus gate voltage. Also, trans-conductance curves were studied by using different widths of GNR as a control parameter to have a high resolution. For all simulations, the length of GNR was fixed at 100 nm (as a channel in FET biosensor) at room temperature. Note that according to Eq. 9, conductance depends only Fermi energy ($E_f$) (as a function of gate voltage) and energy gap ($E_g$) and we know that GNR width affects on energy gap inversely.

3.1.1. The metallic armchair GNR ($N_a = 3m + 2$ family)

In this family, first, second and third subband indexes are equal to $P_1 = 2m + 2$, $P_2 = 2m + 3$ and $P_3 = 2m + 1$ respectively. For the first subband index ($P_1$), the energy gap is zero (as for Eq. 3), this means that the two points of the uppermost valence band and the lowest conduction band meet each others at Dirac point that leads to the metallic behavior for this family. Behavior of these three subband indexes ($p_1$, $p_2$ and $p_3$) is shown in Fig. 2. As we can see in this figure, the sharp deep in $p_1$, that has the most contribution in conductance, corresponds to the zero energy gap whereas by increasing the energy gap (as...
Figure 2: Conductance versus gate voltage for subband indexes $p_1$ (blue line), $p_2$ (black line) and $p_3$ (red line) for (a) Metallic armchair GNR ($N_a = 3m + 2$ family), (b) Semiconductor armchair GNR ($N_a = 3m+1$ family) and (c) Semiconductor armchair GNR ($N_a = 3m$ family) respectively. Here we fixed the width of GNRs at 4nm for all families of armchair GNRs.

happened for $p_2$ and $p_3$) the conductance quickly decreases to gate voltage. In Fig. 3, conductance versus gate voltage by using different GNR widths was investigated for $3m + 2$ family. As seen in the Fig. 3, the GNR width less than 4nm does not provide an acceptable resolution at low gate voltages that leads to limit the sensitivity of FET biosensor. In this figure there are the same trans-conductance curves with respect to $p_1$ for all widths which refer to the ineffectiveness of width changes to control on biosensor sensitivity, because of zero band energy.
3.1.2. The semiconductor armchair GNR ($N_a = 3m + 1$ family)

This family with subband indexes; $P_1 = 2m + 1$, $P_2 = 2m + 2$ and $P_3 = 2m$ has semiconductor behavior due to existence of the energy gap. Contribution of these first three subband indexes versus the conductance show in Fig. 2b. Here can be seen, the first subband index ($P_1$) has the most contribution in conductance while for second and third subband indexes ($P_2$ and $P_3$) conductance are zero with respect to the low range of gate voltage ($-0.4v < V_g < 0.4v$). Conductance was shown by using different widths of semiconductor armchair GNR for $P_1$ depict in Fig. 3b from 4nm to 40nm. Here a large energy gap appeared for the lowest GNR width ($W = 4nm$) in region of gate voltage from $-0.2v$ to $0.2v$ which is correspond to owns the highest sensitivity in this range of GNR width. As shown in this Figure, by increasing the GNR width from 4nm to 40nm (as a result of energy gap reduction), the sharp deep increase in conductance curve versus gate voltage. This means that the resolution which directly relates to FET biosensor sensitivity, increases to detect of Dirac point shift at low gate voltages.

3.1.3. The semiconductor armchair GNR ($N_a = 3m$ family)

First three subband indexes for this family of semiconductor armchair GNR are defined as $P_1 = 2m + 1$, $P_2 = 2m$ and $P_3 = 2m + 2$. Fig. 2c displays the conductance versus gate voltage for each subband index. It is visible that conductance versus low range of gate voltage decrease for $P_2$ and $P_3$ while the most portion in trans-conductance curves is corresponded to with $P_1$. Fig. 3c shows the effects of conductance versus gate voltage for different GNR widths. Also, the energy gap decreases while the width of GNRs increase from 4nm to 40nm. This refers to a decrease in sensitivity of the FET transistor as biosensor in showing a small shift in Dirac point at low gate voltages. Similar to previous family ($3m + 1$), a large energy gap appears in the region from $-0.2v$ to $0.2v$ for the lowest width of GNR ($W = 4nm$) in owns the highest sensitivity. Thus, by increasing width in range of 4 – 40nm the energy gap decreased from 0.285eV to 0.030eV for semiconductor armchair GNR, and also the sensitivity range of
Figure 3: Conductance versus gate voltage for different widths of (a) Metallic armchair GNR ($N_a = 3m + 2$ family), (b) Semiconductor armchair GNR ($N_a = 3m + 1$ family) and (c) Semiconductor armchair GNR ($N_a = 3m$ family). Different widths of ribbons are 4nm, 8nm, 15nm, 20nm, 30nm and 40nm.

Gate voltage for this device was about $-0.2v < V_g < 0.2v$ (for an un-doped GNR as a FET channel).

By rising the energy gap, sharp deep of the trans-conductance curve turns into a wide one indicating a noticeable fall off in the transistor sensitivity. As we can see, the resolution and sensitivity of GNR-FET can be controlled by using the nanoribbon width as a parameter to detect a small Dirac point shift at low gate voltages. It is noteworthy to mention that first subband index in all families of perfect armchair GNR had the most contribution in conductance.

Table 1 shows a comparison between theoretical and experimental results of energy gap [29] with respect to the perfect semiconductor armchair GNR for
Table 1: A comparison between calculated band gap energy \( E_{g(Theo)} \) and experimental results \( E_{g(Exp)} \) \[29\] with respect to different widths of perfect semiconductor armchair GNR.

<table>
<thead>
<tr>
<th>Family Structure</th>
<th>Width (nm)</th>
<th>( E_{g(Exp)} ) (eV)</th>
<th>( E_{g(Theo)} ) (eV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>( 3m + 1 )</td>
<td>2.1</td>
<td>0.549</td>
<td>0.522</td>
</tr>
<tr>
<td>( 3m + 1 )</td>
<td>1.6</td>
<td>0.675</td>
<td>0.669</td>
</tr>
<tr>
<td>( 3m + 1 )</td>
<td>1.1</td>
<td>0.874</td>
<td>0.907</td>
</tr>
<tr>
<td>( 3m )</td>
<td>3.2</td>
<td>0.573</td>
<td>0.356</td>
</tr>
<tr>
<td>( 3m )</td>
<td>2.7</td>
<td>0.661</td>
<td>0.420</td>
</tr>
<tr>
<td>( 3m )</td>
<td>1.7</td>
<td>0.953</td>
<td>0.623</td>
</tr>
<tr>
<td>( 3m )</td>
<td>1.4</td>
<td>1.220</td>
<td>0.767</td>
</tr>
</tbody>
</table>

different widths of \( 3m \) and \( 3m + 1 \) family. From the simulations, energy gap increased as 0.522eV, 0.669eV and 0.907eV for the widths 2.1nm, 1.6nm and 1.1nm of \( 3m + 1 \) family and 0.356eV, 0.420eV, 0.623eV and 0.767eV for widths 3.2nm, 2.7nm, 1.7nm and 1.4nm of \( 3m \) family for semiconductor armchair GNR respectively. So, the energy gap increases as a result of the reduction in GNR width. In addition, for armchair GNR as perfect the \( 3m \) family almost has the larger energy gap than the \( 3m + 1 \), while \( 3m + 2 \) family has a zero gap energy for all \( m \)’s which this is due to a larger width of \( 3m + 1 \) family than \( 3m \) family for the same \( m \). As an example for \( m=15 \), energy gap is equal to 0.22eV, 0.21eV and 0eV respect to \( 3m \), \( 3m + 1 \) and \( 3m + 2 \) family, respectively. Let’s remind again \( m \) is related to \( W \) as Eq. \[4\]. These results \( (E_{g(3m)} \geq E_{g(3m+1)} > E_{g(3m+2)} = 0) \) also have been proven in experimental work \[14\].

3.2. GNR Field Effect Transistor Biosensor

Now, effects of charged lipid membranes coated on GNR (a typical biomimetic membrane) and different concentrations of a type specific biomolecule (Magainin 2 peptides) are evaluated on GNR’s Dirac point shift and energy gap. Simulations are compared with an experimental data \[30\] to calibrate our device as a biosensor. The effects of concentration on thickness of the biomimetic mem-
brane investigate as biomolecules-membrane interaction. Here, we consider the semiconductor armchair GNR \((N_a = 3m\) family) to working as the FET channel while the width of ribbons is a control parameter to adapt the analytical conductance model with experimental results (in this case GNR width was fixed about \(30\) nm), while drain-source voltage \((V_{ds})\) is fixed at \(0.1v\).

Results comparison between simulations and experimental data on conductance to gate voltage changes are shown for bare GNR (Fig. 4(a)) with \(R^2 = 0.997\) (the coefficient of determination), GNR coated with the neutral membrane (Fig. 4(b)), GNR coated with negatively (Fig. 4(c)) and positively charged membranes (Fig. 4(d)) at \(V_{ds} = 0.1v\). As a result of this system calibration, we can see in Fig. 4 the shift in minimum conductivity of GNR \((V_{Dirac})\) with respect to \(V_g = 0v\) about \(0.12v\) and \(0.28v\) to positive gate voltage direction for neutral (with \(R^2 = 0.996\)) and negatively charged membrane (with \(R^2 = 0.996\)) respectively, and also about \(0.1v\) shift to negative gate voltage direction for the positively charged membrane (with \(R^2 = 0.991\)). So the Fermi energy is just a gate voltage dependent parameter (Eq. 9) for an undoped GNR while by introducing a charged biological molecules to the system, electrical charge distribution on GNR is perturbed. This will induce a voltage in the system. Therefore, the induced voltage will modify Eq. 9 as a result of altering the Fermi energy as

\[
\tilde{E}_f = E_f + Q(V_{ds} + V'),
\]

where \(Q\) is the electrical charge of carriers and \(V'\) is induction voltage as a result of assembling impurities on the GNR surface (for an un-doped GNR, \(V' = 0\)).

Now, the electrolyte-gate responses of GNR-FET biosensor is considered when different concentration of Magainin 2 peptides added in \(10mM\) NaF as electrolyte-gate. Comparison between simulation results and experimental data are shown in Fig. 5. Here two different concentrations of Magainin 2 peptides \((0.01nM \ and \ 1\mu M)\) is added to electrolyte solution. Concentration changes were investigated for bare GNR in Fig. 5(a) (with \(R^2 = 0.985\) for \(0.01nM\) peptides and \(R^2 = 0.982\) for \(1\mu M\) peptides) and GNR coated with a negatively
Figure 4: Conductance curves versus gate voltage in comparing between experimental data and simulation results for semiconductor armchair GNR ($N_a = 3m$ family) at $V_{ds} = 0.1$V for (a) un-doped GNR (bare GNR), (b) GNR coated with a neutral membrane ((lipid composition 1-palmitoyl-2-oleoyl-sn-glycero-3-phosphocholine (POPC)), (c) GNR coated with a negatively charged membrane ((POPC)/1-palmitoyl-2-oleoyl-sn-glycero-3-[phosphorac-(1-glycerol)] (POPG) (2/1)) and (d) GNR coated with a positively charged membrane (lipid composition 1,2-dioleoyl-sn-glycero-3-phosphocholine (DOPC)/1,2-dioleoyl-3-trimethylammonium-propane (DOTAP) (2/1). We fixed the width of GNR at 30nm.
Figure 5: Electrolyte-gate responses of GNR to add different concentrations of Magainin 2 peptides in electrolyte solution. (a) Compare transfer curves between experimental data and simulation for bare GNR with 0.01nM and 1µM Magainin 2 peptides and (b) Compare trans-conductance curves between experimental data and simulation for GNR coated with gram-negative bacteria biomimetic membrane with 0.01nM and 1µM of Magainin 2 peptides. Here the width of GNR fixed at 30nm.

charged membrane (gram-negative bacteria biomimetic membrane) in Fig. 5 (with $R^2 = 0.989$ for 0.01nM peptides and $R^2 = 0.912$ for 1µM peptides), respectively. For GNR coated with negatively charged membrane, $V_{\text{Dirac}}$ shifted to negative gate voltage direction about 0.2v and also energy gap increased as a result of increase the concentration from 0.01nM to 1µM. For this case, there was a decrease in energy gap by decreasing thickness of the membrane coated on GNR (from 5nm to 3nm) as a result of interaction between Magainin 2 peptides and negatively charged membrane, which this led to a decrease in thickness by increasing the concentration. For GNR coated with a negatively charged membrane at the same concentration of Magainin 2 peptides, the effects of both change concentration and membrane thickness have to consider on charge carriers of GNR due to the interaction between Magainin 2 peptides and the negatively charged membrane. Here, the simulation results were compared between bare GNR (zero thickness of the membrane) and GNR coated with the negative membrane (5nm and 3nm thickness of the membranes) according to this experimental work for comparing effects of thickness on energy gap. As
seen here, the shift in Dirac point for bare GNR (Fig. 5a) is much smaller shift in $V_{\text{Dirac}}$ compared to that GNR coated with membrane (as seen in Fig. 5b), when the same concentration were added. So, for this system as a hybrid device (GNR coated with a membrane), there is a high sensitivity. The simulations show that the energy gap increased by increasing the electrical charge of biomolecules, and it decreased by decreasing the thickness of the membrane deposited on GNR. Also, the Dirac point shifted to positive direction on the gate voltage axis, for impurities of negative charges (by p-doping of GNR), while the same simulations indicate that the Dirac point shifted to negative direction on the same axis, for impurities with positive charges (by n-doping of GNR).

As we can see, the shift of Dirac point to negative direction on the gate voltage axis can be explained by the adsorption of biomolecules with positive charges. This result is in agreement with the report of Chen et al. [31] that used an electrochemical-gating approach to study the charge transport of single layer graphene in transistors. They showed that, the Dirac point shifted to negative direction on the gate voltage axis as the result of increasing the concentrations of positive charges. Li et al. [32] fabricated a graphene-based FET with tunable sensitivity to sensitive and selective detection of mercury (II) ($Hg^{+2}$) ions in solution. Their results showed that, by increasing the concentration of $Hg^{+2}$ ions, Dirac point shifted to positive direction on the gate voltage axis, while the adsorption of anions on graphene surface the Dirac point shifted to negative direction on the same axis.

4. Conclusions

In the current work, for the first time, we have shown that the control of GNR-FET resolution can be achieved by managing the GNR width to detect low concentrations at low gate voltage. The results showed that the Fermi energy was just a gate voltage dependent parameter for an un-doped GNR while the perturbation of electrical charge distribution introduced by biological molecules
will induce a voltage in the system, which will modify the Fermi energy in an un-doped GNR biosensor. According to the simulations, energy gap increased by increasing the concentration, while it decreased by decreasing thickness of the membrane coated on GNR. And also, the results show that the $V_{\text{Dirac}}$ shifted towards the positive and negative direction on the gate voltage axis for negatively and positively charged impurities, respectively.

This work highlights the promising potentials of GNR in nanoelectronics and bioelectronics by managing the GNR width to control the sensitivity of biosensors which is strong evidence for applying biosensors to detect charged biomolecules with low concentrations. The theoretical findings here, provide an important step toward developing GNR devices as biosensors with high resolution and sensitivity. This discussions help researchers to better understand the effects of the GNR width in conductance and resolution to design powerful and flexible devices with high sensitivity that will have solid significance in biosensors devices in the future.

Now we are working on poly(3,4-ethylenedioxythiophene):polystyrene sulfonate (PEDOT:PSS) conductance to develop the conductance model for both GNR and PEDOT:PSS in hybrid transistors to detect some biological molecules in organic electrochemical transistors (OECTs) to detect concentrations of some biomolecules with both electrical and chemical properties. This will be a new method to detect redox active biomolecules by GNR-biotransistor based on PEDOT:PSS as hybrid devices with high resolution.

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• The perturbation of electrical charge distribution introduced by biological molecules will induce a voltage in the system, which will modify the Fermi energy in an un-doped GNR biosensors.

• The resolution and sensitivity of FET-GNR biosensors can be controlled by using the nanoribbon width as a parameter to detect a small shift of Dirac point at low gate voltages.

• Energy gap increased by increasing the electrical charge of biomolecules while it decreased by decreasing thickness of the membrane coated on GNR.
Author contributions

Use this form to specify the contribution of each author of your manuscript. A distinction is made between five types of contributions: Conceived and designed the analysis; Collected the data; Contributed data or analysis tools; Performed the analysis; Wrote the paper.

For each author of your manuscript, please indicate the types of contributions the author has made. An author may have made more than one type of contribution. Optionally, for each contribution type, you may specify the contribution of an author in more detail by providing a one-sentence statement in which the contribution is summarized. In the case of an author who contributed to performing the analysis, the author’s contribution for instance could be specified in more detail as ‘Performed the computer simulations’, ‘Performed the statistical analysis’, or ‘Performed the text mining analysis’.

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**Manuscript title:** Design of Field Effect Transistor Biosensor based on Graphene Nanoribbons with High Resolution

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Declaration of interests

☒ The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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