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Investigation and comparison of graphene nanoribbon and carbon nanotube based SARS-CoV-2 detection sensors: An ab initio study

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ARTICLE INFO
Keywords:
SARS-CoV-2
Sensor
Graphene nanoribbon
Carbon nanotube
PBASE linker
Antibody

ABSTRACT
The rapid detection of SARS-CoV-2, the pathogen of the Covid-19 pandemic, is obviously of great importance for stopping the spread of the virus by detecting infected individuals. Here, we report the ab initio analysis results of graphene nanoribbon (GNR) and carbon nanotube (CNT) based SARS-CoV-2 detection sensors which are experimentally demonstrated in the literature. The investigated structures are the realistic molecular models of the sensors that are employing 1-pyrenebutyric acid N-hydroxysuccinimide ester as the antibody linker. Density functional theory in conjunction with non-equilibrium Green’s function formalism (DFT-NEGF) is used to obtain the transmission spectra, current-voltage and resistance-voltage characteristics of the sensors before and after the attachment of the SARS-CoV-2 spike protein. The operation mechanism of the GNR and CNT based SARS-CoV-2 sensors are exposed using the transmission spectrum analysis. Moreover, it is observed that GNR based sensor has more definitive detection characteristics compared to its CNT based counterpart.

1. Introduction
The severe acute respiratory syndrome is the pathogen of the coronavirus disease 2019 (Covid-19), which is ongoing as a global pandemic since the beginning of 2020. The identification of the infected individuals is at the center of the confronting strategies. The standard method of testing for Covid-19 is the employment of the polymerase chain reaction (PCR) tests [1–3]. The diagnosis duration using PCR tests has a median of 3 h [4,5]. Considering that the PCR tests are generally carried in centralized laboratories, the diagnosis time often extend during the peaks of the pandemic when fast results are of crucial importance. Therefore, point-of-care (PoC) rapid tests, which carry the potential for much faster diagnosis results, are under consideration [6].

There are several rapid test methods presented in the literature. In one of these studies, a biosensor utilizing membrane-engineered mammalian cells with human chimeric spike antibody is demonstrated where the cellular bioelectric properties are used as the indicator [7]. In another study, the change of the optical properties of functionalized single-walled carbon nanotubes (SWCNTs) by the attachment of the SARS-CoV-2 spike protein is employed for the detection [8]. A similar optical sensor was implemented for the detection of the SARS-CoV-2 existence in air samples [9]. It is demonstrated that three different optical detection techniques can be used for the SARS-CoV-2 detection with various advantages and disadvantages for each method [10]. A sensor-modified electrode chip using DNA linker is also shown to give fast results [11]. In another study, it is shown that electrochemical sensors with cotton-tip can be used to sense SARS-CoV-2 antigen [12]. In another approach, the differences of the magnetic field behaviour of functionalized magnetic nanoparticles are employed for the detection of Covid-19 pathogen [13]. A clustered regularly interspaced short palindromic repeats (CRISPR)/Cas9 endonuclease dead (dCas9) system is also demonstrated to operate as a colorimetric detection sensor for SARS-CoV-2 [14]. In another work, a paper-based electrochemical sensor using gold nanoparticles is employed for the detection [15]. Detection of SARS-CoV-2 from exhaled breathe is clearly of high importance for testing and in Ref. [16], such a sensor using nanomaterial based hybrid sensor arrays is presented. A similar electrochemical sensor with a detection limit of 15 fM is realized in Ref. [17]. In Ref. [18], a CPE–HT18C6(Ag)/chitosan/SiQDs@PAMAM/probe sequence for the voltammetric detection of SARS-CoV-2 RNA-dependent RNA polymerase (RdRp) is implemented which has a relative standard deviation for single electrode repeatability having a value less than 2.7%. In another work, the detection of SARS-CoV-2 antibodies using capacitive immunosensing assay is reported where commercially

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available CR3022 antibody is utilized to test the system [19]. Similarly, a sensitive sensor is demonstrated for the detection of immunoglobulin using a paper-based sensor without the need of specific antibody requirements [20]. In Ref. [21], a colorimetric assay based on gold nanoparticles is reported in which the test results are made available in 10 min. In a different approach, plasmonic optical fibers and molecularly imprinted nano polymers are used for the implementation of a SARS-CoV-2 sensor which also operates in the 10 min window [22]. An acoustical slot mode sensor which also completes the test in the 10 min time window is presented in Ref. [23] where slot wave in an acoustical delay line was employed for the detection. An electrochemical immunonoassay using magnetic beads and carbon black nanomaterial is implemented in Ref. [24] in which the detection limits for the spike protein and nucleocapsid protein of SARS-CoV-2 is reported to be 19 ng/mL and 8 ng/mL, respectively, with the test time of 30 min. The cost of rapid testing is obviously of high importance as millions of tests have to be carried in peak times. Considering the cost-effectiveness, an electrochemical sensor having 1.68 ng/mL sensitivity which can be manufactured with a process similar to that of glucose test strips, is proposed in Ref. [25]. Detection of SARS-CoV-2 specific antibodies using an opto-microfluidic sensing platform employing gold nanoelectrodes is also demonstrated to detect the SARS-CoV-2 antibodies using the local surface plasmon resonance method [26]. In another study, the operating mechanism of a surface plasmon resonance based sensor using gold nanorods is explained in detail [27]. In a similar study, phase-modulation plasmonic sensors are used to detect SARS-CoV-2 spike protein with a high performance detection sensitivity of 8.4069 × 10^4/RIU (refractive index unit) [28]. A practical low-cost rapid electrochemical detection system based on a disposable Si MOSFET sensor unit employing glass slide and gold metal lines functionalized with SARS-CoV-2 spike antibody is shown to detect SARS-CoV-2 spike protein concentrations at the 7 fm sensitivity limit [29]. In another work, microfluidic ELISA (enzyme-linked immunosorbent assay) based sensor employing lab-on-chip platform is implemented for the detection of SARS-CoV-2 antibodies with a high sensitivity and selectivity [30].

There are various linkers used to functionalize nano materials for use as sensors in the literature. 1-Pyrenenbutanoic acid succinimidyl ester (PBASE) is one of these linkers frequently used for functionalizing both carbon nanotubes (CNTs) and graphene nanoribbons (GNRs) [31]. The studies regarding the utilization of PBASE for the protein linking to the multi-walled CNTs (MWCNTs) are demonstrated back in 2006 [32]. Non-covalent functionalization of graphene based field effect transistors (FETs) and their use as biosensors are demonstrated in another study in 2011 [33]. Similarly, PBASE functionalized graphene sheets are used as prostate specific antigen (PSA) detection sensor [34]. PBASE functionalized graphene based FET sensors are also used for the sensitive detection of immunoglobulin levels [35]. A DNA biosensor is implemented using graphene-PBASE nanocomposite which was prepared by the ultrasonication method as demonstrated in Ref. [36]. In another study, a glucose sensor is produced on graphene oxide (GO) functionalized with 1-pyrenenbutyric acid-N-hydroxysuccinimide ester (PANHS) showing a high sensitivity up to the value of 40.5 ± 0.4 mA M⁻¹ cm⁻² [37]. A high-performance bioelectronic tongue is implemented employing CNT with PBASE functionalization which displays a high selectivity among different sweeteners [38]. In another study, pyrenenbutyric acid is used to link MWCNTs and 2-Deoxyribose-5-phosphate Aldolase (DERA) successfully [39]. A low-cost glucose sensor based on the thiol (DERA)functionalized graphene sheet is detected on graphene powder are reported in Ref. [40]. An escherichia coli (E. coli) sensor for food safety applications is implemented using antibodies attached to graphene FETs with the help of PBASE is demonstrated in another study [41]. A rapid and label-free sensor for the detection of breast cancer cells in the blood is produced using IgG antibodies attached to the CNTs using PBASE [42]. In another study, a DNA sensor was implemented using PBASE as a linker on the graphene flakes [43]. In Ref. [44] it is demonstrated that PBASE linked carboxylated MWCNT sensors provide superior sensitivity compared to isolated carboxylated MWCNT sensors for the detection of serum insulin levels. Another study also reported sensitivity enhancements for the single stranded RNA detection using PBASE as a linker for the graphene/zinc oxide base [45]. A colorimetric sensor for histamine is developed using PBASE as linker to the carbon nanoparticles for semi-quantitative results for the histamine range of 0–600 µg/mL [46]. In another study, PBASE linker is used for the attachment of formaldehyde dehydrogenase (FDH) on MWCNTs for the implementation of a sensor for the detection of formaldehyde in diluted urine [47]. Another E. coli sensor is developed using the PBASE molecule as a linker between graphene FETs and E. coli antibodies with different phosphate buffered saline (PBS) buffers [48]. In another study, a genosensor for the detection of exon-19 mutation in epidermal growth factor receptor (EGFR) is implemented on a zirconia-graphene nanocomposite using the PBASE as the linker [49].

Considering the wide usage of PBASE as linker to nano materials, there are several studies regarding their attachment in the literature. One of the first papers explaining the π-stacking of the pyrenyl group demonstrates the sensing of biomolecules with SWCNTs [50]. The π-π attachment geometries of pyrene molecules and CNTs are investigated by both molecular dynamics (MD) simulations and f fluorescence spectroscopy in Ref. [51]. Similarly, the adsorption of 1-pyrenenbutyric acid (PBA) to SWCNTs are experimentally studied in Ref. [52] where the adsorption parameters are computed. The molecular structure of PBASE and the complexes formed with SWCNTs and PBASE are obtained using density functional theory (DFT) in Ref. [53]. The attachment of PBASE between graphene layers and their geometries are reported in Ref. [54]. The π-stacking interactions of PBASE and SWCNTs are also given in Ref. [55] using both ab initio and experimental discussions. The adsorption of PBA on graphene layers is investigated via ab initio calculations, Raman spectroscopy and spectroscopic ellipsometry in Ref. [56]. The attachment of pyrene ligands on various carbon nanomaterials including SWCNTs and graphene are studied experimentally by high-resolution mass spectrometry, NMR, UV–Vis, X-ray crystallography and steady-state fluorescence spectroscopy for the geometric explanation of these systems in Ref. [57]. The structural properties of pyrene-linked polyamide SWCNTs and MWCNTs are studied using MD simulations in Ref. [58]. A pyrene functionalized armchair graphene nanoribbon field effect sensor is developed in Ref. [59] for use in the uridine diphosphate glucose detection. Due to the established information on graphene and CNT sensors functionalized with PBASE, several detectors for SARS-CoV-2 have also been developed using these complexes. In one of these studies, a graphene FET sensor is realized by the attachment of SARS-CoV-2 antibodies on the graphene channel using PBASE as a linker where a detection sensitivity of 100 fg/mL is achieved [60]. In another study, PBASE is again used as a linker between the graphene and SARS-CoV-2 antibodies to detect spike protein in blood or saliva samples [61]. A graphene sensor for the detection of SARS-CoV-2 nucleic acid or spike proteins is developed in Ref. [62] where PBASE is employed as a linker between the graphene layer and the ss-DNA probes. It is shown that the slope of the linear current-voltage (I–V) characteristics of the graphene layer changes by the concentration of the RNA-dependent RNA polymerase (RdRp) [62]. In Ref. [63], a proof of concept immunosensor for the detection of SARS-CoV-2 spike protein by the spike-specific antibodies attached on graphene using PBASE is reported in which the characteristics of the sensor is given. In another work, two types of sensors implemented by graphene sheets and CNTs linked to antibodies using PBASE are tested by simulation [64].

In this work, SARS-CoV-2 detection sensors using GNRs and CNTs are investigated using ab initio methods following the experimental realizations demonstrated in Refs. [60–63]. DFT in conjunction with non-equilibrium Green’s function formalism (NEGF) is used to obtain the transmission spectra and I–V characteristics of equivalent GNR and CNT based SARS-CoV-2 sensors. The utilization of CNTs and GNRs as bases for sensors and their analysis with the transmission spectrum concept is widely studied in the literature [65–67]. The transmission
spectra of i) the bare GNR (CNT) 2-probe devices, ii) PBASE functionalized GNR (CNT) 2-probe devices, iii) PBASE functionalized GNR (CNT) 2-probe devices with the SARS-CoV-2 antibody complexes (sensors) and finally iv) the sensors with the attached SARS-CoV-2 spike protein are obtained step-by-step. The I–V and resistance-voltage (R–V) curves of the investigated GNR and CNT based sensors are then obtained for comparison. It is worth noting that the analysed structures consist of thousands of atoms without any fragmentation in simulations for achieving the best possible accuracy in the DFT-NEGF computations. The results explain the mechanism by which antibody-functionalized GNRs and CNTs can be used for the detection of SARS-CoV-2. It is also exposed that the GNR-PBASE-antibody based sensor have better detection characteristics compared to its equivalent CNT-PBASE-antibody based counterpart.

2. Material and methods

The sensors considered in this study are based on equivalent conducting GNR and CNT 2-probe devices. The GNR sensor employs an 18-atoms wide zigzag GNR while the CNT is a (5,5) armchair CNT. These dimensions are selected so that the unit cells of conducting GNR and CNT have the same number of atoms. These GNR and CNT types belong to asymmetric zigzag and armchair groups, respectively, which have conductive behaviours [68,69]. The equivalent metallic GNR and CNT 2-probe devices considered in this study are shown in Fig. 1 where both structures consist of 40 periods of unit cells and 800 atoms.

The metallic GNR and CNT 2-probe devices are functionalized using PBASE as the first step. The chemical formula of the PBASE is C_{24}H_{19}NO_{4} [53,70]. The optimized molecular structure of the PBASE and metallic GNR and CNT devices with PBASE molecules attached are shown in Figs. 2 and 3, respectively.

It is demonstrated in Refs. [71,72] that the human neutralizing antibody CR3022 binds to the SARS-CoV-2 receptor binding domain (RBD) with a K_{D} of 6.3 nM. In addition CR3022, which is commercially available [73], is an antibody used in SARS-CoV-2 sensors as shown in previous studies [74–76]. Therefore, CR3022 antibody is bound to the PBASE linker to form the GNR and CNT based SARS-CoV-2 sensors as the next step as shown in Fig. 4. The sensors shown in Fig. 4 are now ready to be used to detect SARS-CoV-2 spike protein. The binding geometry of the CR3022 and the SARS-CoV-2 spike protein are taken from Refs. [71,77] which is obtained using cryo-EM molecular imaging. The sensor structures with the SARS-CoV-2 spike proteins bound are shown in Fig. 5 as the last step. The number of atoms of each molecule and 2-probe devices are given in Table 1. It is worth noting that these structures are simulated without any fragmentation to achieve accuracy.

Density functional theory (DFT) [78] in conjunction with non-equilibrium Green’s function formalism (NEGF) [79] is used in this study for obtaining the current-voltage (I–V) and related parameters of
the sensors. QuantumATK® software is used for performing the simulations [80]. All of the structures shown in Figs. 1–5 are optimized until the forces on each atom are less than 0.05 eV/Å in QuantumATK® as the first step. Then, the actual DFT-NEGF computations are performed with the following parameters: the density mesh cutoff is 680 eV, exchange-correlation functional is generalized gradient approximation (GGA) [81], iteration algorithm is pulay mixer with 20 history steps. The DFT computations include van der Waals dispersion corrections as used in various studies in the literature [82–85]. It is worth noting that the number of atoms is on the order of thousands for the sensor as shown in Table 1 and the DFT-NEGF computations of these large molecules are performed in QuantumATK® with the utilization of intelligent computational load distribution among processors on the simulation servers having 128 GB of RAM each.

### 3. Results and discussion

The DFT-NEGF simulations are performed until the convergences are reached and then the transmission spectra of the simulated structures between [0 eV–1 eV] energy range with 251 data points are obtained as the first and main outputs. The transmission spectra of the bare 2-probe GNR device, 2-probe GNR device with the PBASE linker, 2-probe GNR device with PBASE and the CR3022 antibody complex (sensor) and the 2-probe GNR device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached are computed as given in Fig. 6. As it can be observed from Figs. 6 and 7, the transmissions of the bare 2-probe GNR and CNT devices have the values of 1 and 2 in the whole energy range due to the number of transmission channels as expected [86,87]. Moreover, the attachments of the PBASE linker, CR3022 antibody and SARS-CoV-2 spike protein RBD dramatically change the transmission spectra of both GNR and CNT 2-probe devices making them suitable for use as SARS-CoV-2 sensors. In addition, the differences of the I–V characteristics are required for a better evaluation and comparison of these sensors. For this aim, the current values are calculated using the Landauer-Buttiker formula which is shown in Eq. (1) [88].

$$I = \frac{2q}{\hbar} \int T(E) \left[f_L(E) - f_R(E)\right] dE$$

In Eq. (1), $q$ is the elementary charge, $\hbar$ is Planck’s constant, $T(E)$ is the transmission spectrum, $f_L(E)$ and $f_R(E)$ are the Fermi-Dirac distribution functions of the left and right electrodes, $\mu$ is the equilibrium electrochemical potential and $V$ is the applied potential difference [81]. The current values of the simulated structures are calculated using the Landauer-Buttiker formula for 251 voltage points and plotted on the same axes in Figs. 8 and 9 for the GNR and CNT sensors, respectively.

The attachment of the PBASE, CR3022 antibody and the SARS-CoV-2 spike complex clearly affects the I–V characteristics of the sensors as observed from Figs. 8 and 9. The change of the I–V curve of the CNT device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike RBD attached is computed as given in Fig. 7. As it can be observed from Figs. 6 and 7, the transmissions of the bare 2-probe GNR and CNT devices have the values of 1 and 2 in the whole energy range due to the number of transmission channels as expected [86,87]. Moreover, the attachments of the PBASE linker, CR3022 antibody and SARS-CoV-2 spike protein RBD dramatically change the transmission spectra of both GNR and CNT 2-probe devices making them suitable for use as SARS-CoV-2 sensors. In addition, the differences of the I–V characteristics are required for a better evaluation and comparison of these sensors. For this aim, the current values are calculated using the Landauer-Buttiker formula which is shown in Eq. (1) [88].

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The attachment of the PBASE, CR3022 antibody and the SARS-CoV-2 spike complex clearly affects the I–V characteristics of the sensors as observed from Figs. 8 and 9. The change of the I–V curve of the CNT device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached is computed as given in Fig. 7. As it can be observed from Figs. 6 and 7, the transmissions of the bare 2-probe GNR and CNT devices have the values of 1 and 2 in the whole energy range due to the number of transmission channels as expected [86,87]. Moreover, the attachments of the PBASE linker, CR3022 antibody and SARS-CoV-2 spike protein RBD dramatically change the transmission spectra of both GNR and CNT 2-probe devices making them suitable for use as SARS-CoV-2 sensors. In addition, the differences of the I–V characteristics are required for a better evaluation and comparison of these sensors. For this aim, the current values are calculated using the Landauer-Buttiker formula which is shown in Eq. (1) [88].

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The attachment of the PBASE, CR3022 antibody and the SARS-CoV-2 spike complex clearly affects the I–V characteristics of the sensors as observed from Figs. 8 and 9. The change of the I–V curve of the CNT

### Table 1

| Structure                                      | Number of atoms |
|-----------------------------------------------|-----------------|
| Metallic GNR 2-probe device                   | 800             |
| Metallic CNT 2-probe device                   | 800             |
| PBASE molecule                                | 48              |
| Metallic GNR 2-probe device with PBASE        | 848             |
| Metallic CNT 2-probe device with PBASE        | 848             |
| Metallic GNR 2-probe device with PBASE and CR3022 complex (sensor) | 7495 |
| Metallic CNT 2-probe device with PBASE and CR3022 complex (sensor) | 7495 |
| Metallic GNR 2-probe device with PBASE and CR3022 complex (sensor) as the SARS-CoV-2 spike protein RBD attached | 10,500 |
| Metallic CNT 2-probe device with PBASE and CR3022 complex (sensor) as the SARS-CoV-2 spike protein RBD attached | 10,500 |

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Fig. 5. GNR (a) and CNT (b) sensors with the SARS-CoV-2 spike protein RBDs attached.

Fig. 6. The transmission spectra of the bare 2-probe GNR device, 2-probe GNR device with the PBASE linker, 2-probe GNR device with PBASE and the CR3022 antibody complex (sensor) and the 2-probe GNR device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached.

Fig. 7. The transmission spectra of the bare 2-probe CNT device, 2-probe CNT device with the PBASE linker, 2-probe CNT device with PBASE and the CR3022 antibody complex (sensor) and the 2-probe CNT device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached.
sensor after the attachment of the PBASE molecule is greater than that of the GNR sensor and the reason of this is the difference of the changes of the transmission spectra shown in Figs. 6 and 7. The transmission spectrum of the CNT sensor changes dramatically after the addition of the PBASE linker molecule as can be seen from Fig. 7 and this is reflected in the change of the I–V characteristics of Fig. 9. In comparison, the change of the transmission spectrum of the GNR based sensor is smaller after the addition of the PBASE molecule in Fig. 6 and this effect is seen as the small I–V differences of Fig. 8.

The I–V characteristics of the sensors (GNR, CNT + PBASE + CR3022) before and after the attachment of the SARS-CoV-2 spike protein is obviously the focus of this work. These I–V curves are obtained for the [0V–0.1 V] voltage range and plotted in Figs. 10 and 11 for the GNR and CNT bases sensors, respectively. The analyses are focused in this voltage range considering the low-voltage trend in microelectronic devices. In addition, the resistance-voltage (R–V) characteristics of the sensors are also computed and given in Figs. 12 and 13. The following results are observed from the I–V and R–V characteristics of the GNR and CNT based sensors before and after the attachment of the SARS-CoV-2 spike proteins: i) the current levels of these sensors are on the order of a few microamperes (μA) which make both of them practical from the electronic current sensing point of view, ii) the resistance values are on the order of a few tens of kiloOhms (kΩ) which can easily be measured and processed by Wheatstone bridge based electronic circuits, iii) comparing Figs. 10 and 11, it is observed that the I–V characteristics of the GNR based sensor displays more definitive changes after the attachment of the SARS-CoV-2 spike protein RBD compared to the CNT based sensor, iv) comparing Figs. 12 and 13, it is observed that the R–V characteristics of the GNR based sensor shows greater variation after the attachment of the SARS-CoV-2 spike protein RBD compared to the CNT based sensor. The I–V or R–V variations of these sensors can be detected using complementary metal-oxide-semiconductor (CMOS) integrated circuits [89,90]. It is worth noting that these differences of the current

Fig. 8. The current-voltage characteristics of the bare 2-probe GNR device, 2-probe GNR device with the PBASE linker, 2-probe GNR device with PBASE and the CR3022 antibody complex (sensor) and the 2-probe GNR device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached.

Fig. 9. The current-voltage characteristics of the bare 2-probe CNT device, 2-probe CNT device with the PBASE linker, 2-probe CNT device with PBASE and the CR3022 antibody complex (sensor) and the 2-probe CNT device with PBASE and CR3022 antibody complex (sensor) having the SARS-CoV-2 spike protein RBD attached.

Fig. 10. The current-voltage characteristics of the GNR based sensor with and without the attachment of the SARS-CoV-2 spike protein RBD.

Fig. 11. The current-voltage characteristics of the CNT based sensor with and without the attachment of the SARS-CoV-2 spike protein RBD.

Fig. 12. The resistance-voltage characteristics of the GNR based sensor with and without the attachment of the SARS-CoV-2 spike protein RBD.
and resistance values have the roots in the changes of the transmission spectra after the attachment of the SARS-CoV-2 spike proteins.

4. Conclusions

First-principles computations of the GNR and CNT based SARS-CoV-2 sensors are reported for the first time in this work. Density functional theory in conjunction with non-equilibrium Green’s function formalism is utilized for obtaining the transmission spectra and current-voltage variations of the considered biosensors. The transmission spectra and the current-voltage characteristics of i) the equivalent bare 2-probe GNR and CNT devices, ii) 2-probe GNR and CNT devices with the PBASE linker, iii) 2-probe GNR and CNT devices with the PBASE linker and the CR3022 antibody complexes (sensors) are obtained individually. As the next step, the transmission spectra, the current-voltage and the resistance-voltage characteristics of these sensors with and without the attachment of the SARS-CoV-2 spike protein RBD are simulated and plotted. It is observed that the attachment of the SARS-CoV-2 spike protein affects the I–V and R–V characteristics for both GNR and CNT based sensors. The reason for the changes in the I–V and R–V characteristics stems from the change in the transmission spectra of the sensors when SARS-CoV-2 spike protein is attached. Moreover, it is observed that the change in the I–V and R–V characteristics of the GNR based sensor after sensing the SARS-CoV-2 spike protein RBD is more distinctive compared to its CNT based counterpart. The results of this study are expected to aid in the understanding of the operation mechanism of the GNR and CNT based SARS-CoV-2 sensors, which are vital in the development of portable, ultra-rapid and ultra-sensitive cell-based biosensor for the direct detection of the SARS-CoV-2 spike protein antigen, Sensors 20 (2020) 3211.

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Data availability

Data will be made available on request.
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