Scientia Iranica F (2017) 24(6), 3531–3535

Sharif University of Technology
Scientia Iranica
Transactions F: Nanotechnology
www.scientiairanica.com

Graphene-based nano bio-sensor: Sensitivity improvement

M. Majdi and D. Fathi*

School of Electrical and Computer Engineering, Tarbiat Modares University (TMU), Tehran, P.O. Box 14115-194, Iran.

Received 8 August 2016; received in revised form 7 October 2016; accepted 4 March 2017

KEYWORDS
Biosensor;
Sensitivity enhancement;
S-parameter;
Surface plasmon resonance;
Graphene.

Abstract. In this paper, we attempt to present a multilayer-structure biosensor. A graphene-based Surface Plasmon Resonance (SPR) biosensor is considered as a structure, which offers enhancement of sensitivity in comparison with the conventional biosensors. The sensitivity improvement caused by the presence of graphene is discussed through monitoring of the biomolecular interactions and following that the Refractive Index (RI) changes. For this purpose, an RI change, which increases during the course of biomolecular interaction, is considered from 1.462 to 1.52. Our numerical results show that the proposed SPR biosensor with a graphene layer can provide a better sensitivity due to the field distribution at the binding region. This paper investigates the Sensitivity Enhancement Factor (SEF) according to the S-parameters and the effects of design parameters such as thicknesses of the gold and graphene layers and the refractive index change during the course of DNA hybridization in this biosensor.

1. Introduction

Surface Plasmon Resonance (SPR) is an optical method based on the excitation of collective free electrons oscillations at the metal-dielectric interface [1]. The SPR biosensors are supported by the Surface Plasmon Polaritron (SPP) waves in order to investigate and detect the biomolecular interactions. In this method, a certain incidence light angle (resonance angle) can be applied to obtain the required SPR condition [1]. Under this condition, momentum matching between the incident photon and the surface plasmon is attained and the Attenuated Total Reflection (ATR) occurs [1]. The resonance angle as a function of the refractive index of sensing medium can be measured as the biosensor response to the biomolecular interactions surrounding it. During the sensing operation, the adsorption of biomolecules to the receptor molecules will produce a local change in the refractive index at the metal-dielectric interface and enables us to detect these biomolecules by measuring the shift of resonance angle [2].

In this paper, the sensitivity enhancement has been investigated by designing a multilayer structure in which a graphene layer has been applied. The proposed multilayer biosensor has been modelled and simulated using the Finite Element Method (FEM) and the Finite Difference Frequency Domain (FDFD) method using CST MICROWAVE STUDIO [3]. The aim of this simulation is the investigation of sensitivity enhancement with the measurement of scattering parameters (S-parameters) with respect to the refractive index occurring during the biomolecular interactions and the thicknesses of gold and graphene layers. The incorporation of graphene layer in the conventional SPR biosensor can enhance the sensitivity. The lack

*. *Corresponding author. Tel.: +98 21 82884973; Fax: +98 21 82884325*
E-mail address: d.fathi@modares.ac.ir (D. Fathi)

doi: 10.24200/sci.2017.4430

of band gap in graphene turns it to a semi-metal different from other semiconductor materials. These unique properties of graphene can be used in future electronic and optoelectronic devices [4]. In this paper, different properties of graphene have been used in order to improve sensitivity of the proposed biosensor [4].

One of the disadvantages of conventional SPR biosensors is the lower limit of detection and their sensitivity is faced with small ligands and low-concentration analytes [1]. To overcome this limitation, several approaches have been proposed. In this paper, in order to resolve this issue, a graphene-based biosensor has been introduced. The improved sensitivity due to the interpolation of a graphene layer into the conventional SPR biosensor comes from the increased adsorption of biomolecules [5]. Moreover, applying a graphene layer modifies the propagation constant of SPP, which will be followed by changing the sensitivity to the refractive index variation [5].

2. Materials and method

The schematic of the proposed SPR biosensor is shown in Figure 1. To excite the surface plasmon, the light is coupled through the flint glass prism with a specific incidence angle.

The base of the prism is covered by a thin gold film with the thickness of 50 nm and a graphene layer on top of the gold layer with the thicknesses of 2 nm. Binding analytes (biotin-streptavidin; DNA hybridization reaction) are considered as a layer with an initial refractive index of 1.462 [6] in an aqueous medium of phosphate buffer solution, the refractive index of which is 1.33 [7].

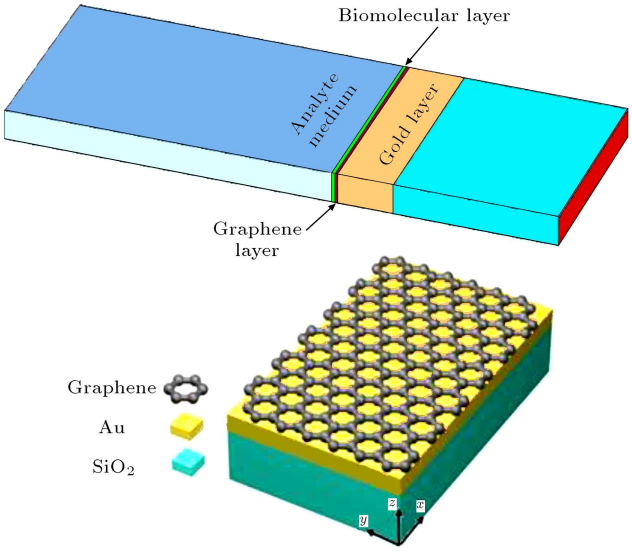


Figure 1. 3-D model of the proposed graphene based biosensor.

In order to find out the best coupling and produce the necessary resonance condition, a TM-polarized plane wave light with the fixed wavelength of 632.8 nm and different incidence angles is coupled into the prism. The optical constants, $\varepsilon = (n, k)$, of glass prism and gold layer are set to (1.75, 0) and (0.178, 3.07), respectively, at $\lambda = 632.8$ nm [8].

It has been shown that graphene has a broad and constant absorption range in the visible spectral region [4]. In this method, a model has been developed for the in-plane optical conductivity of graphene in the visible range. In order to calculate the permittivity, the transmittance, and the reflectance of graphene in the visible range, perturbation Hamiltonian is considered as [4]:

$$H'(t) = -\mu\varepsilon(t), \quad \varepsilon(t) = Re\left[\varepsilon_x e^{-i\omega t}\hat{x}\right], \tag{1}$$

where μ is the electric dipole moment.

By applying the suitable approximation and the simplification of equation, conductivity can be obtained as [4]:

$$\sigma(\omega) = -\frac{ie^2}{\omega A}\sum_k \nu_x^2 \frac{(\rho_{\nu\nu} - \rho_{cc})}{E_{c,k} - E_{\nu,k} - \hbar(\omega + i\eta)}, \tag{2}$$

where A is the cross-section area and ν_x is the velocity operator along x-direction that can be related to the position operator x by the equation of motion.

As we know, the first Brillouin zone of graphene is a hexagon with a unit cell of two atoms. The conduction and valence bands touch each other at the corners of hexagon, which are called the Dirac points. For an intrinsic or lightly doped graphene, the Fermi energy level is around the Dirac points, where the charge carriers experience only a linear dispersion [9]. This linear dispersion is called the Dirac cone since it is described by the relativistic Dirac equation [9]. To calculate the optical properties of graphene in the visible range, one can consider only the Dirac cone if the photon frequency is low compared to the resonance frequency and the Fermi energy is near the Dirac points [4,9].

The direct result of the linear electronic dispersion of Dirac cone is the effective massless fermions, such that the related energy dispersion can be written as [10]:

$$E = \pm\hbar\nu_f|\vec{k}|, \tag{3}$$

where $\hbar$ is the reduced Plank's constant, ν_f is the Fermi velocity in graphene, and $\vec{k}$ is the wave vector. In order to obtain an analytical form of conductivity, we can apply the Dirac cone approximation wherein we assume that conductivity contributes only through the carriers to the Dirac cone. Thus, permittivity, ε, and

the optical conductivity of graphene, σ_0, in the visible range are related by [4]:

$$\varepsilon(\omega) = 5.5\varepsilon_0 + i\frac{\sigma_0}{\omega d}, \tag{4}$$

where ε_0 is the permittivity of vacuum and d is the thickness of graphene. Therefore, the refractive index of graphene in the visible range can obtained [4]:

$$n(\lambda) = 3 + i\frac{5.446}{3}\lambda_0, \tag{5}$$

where λ_0 is the vacuum wavelength.

3. Results and discussion

Our numerical model is based on the biomolecular interaction of DNA (DNA hybridization). As mentioned in the previous section, it is defined as a layer with the refractive index of 1.462. The initial value of 1.462 corresponds to an ssDNA layer with a density of 0.028 g/cm^3, obtained from the ellipsometry measurements. After the hybridization events, the dsDNA layer with a density of 0.061 g/cm^3 corresponding to the refractive index of 1.52 will be obtained [11].

The optical characteristics (e.g., S-parameters) of the biosensor are affected by the biomolecular interactions between ssDNA and cDNA. Measuring the shift of S-parameters is the base of our investigation in this paper.

For achieving the coupling between the plasmon resonance and incident photons, the following condition must be maintained to excite the SPs on the metal-dielectric interface [12]:

$$K_{sp} = K_{ev} = \frac{\omega}{c}\sqrt{\frac{\varepsilon_M\varepsilon_D}{\varepsilon_M + \varepsilon_D}} = \frac{\omega}{c}\sqrt{\varepsilon p}\sin\theta_{\text{res}}, \tag{6}$$

where K_{sp} and K_{ev} are the wave vectors of the surface plasmon and the evanescent wave, respectively, c is the speed of light, ε_M and ε_D are the dielectric constants of metal and dielectric layer, respectively, and θ_{res} is the resonance angle [12]. The S-parameters are complex-valued wavelength-dependent matrices expressing the device characteristics (e.g., the transmission and reflection of electromagnetic energy) using the amount of absorption or transmission. For a two-port device, the S-parameters are defined as [13]:

$$S = \begin{bmatrix} S_{11} & S_{12} \\ S_{21} & S_{22} \end{bmatrix}, \tag{7}$$

where the matrix elements S_{11}, S_{12}, S_{21}, and S_{22} are known as the scattering parameters or the S-parameters. In the meantime, S_{11} and S_{22} represent reflection coefficients, whereas S_{12} and S_{21} are defined as the transmission coefficients. In addition, we have [13]:

$$S_{11} = \sqrt{\frac{\text{Power reflected from 1}}{\text{Power incident on 1}}},$$

$$S_{21} = \sqrt{\frac{\text{Power delivered to 2}}{\text{Power incident on 1}}}. \tag{8}$$

The magnitude of S_{11} ($|S_{11}|$) gives the normalized reflectance. The normalized reflectance for this multilayer SPR is calculated using [1]:

$$S_{11} = \left|\frac{M_{11}}{M_{12}}\right|^2 \exp(i\varphi_{11}), \tag{9}$$

where φ_{11} contains the phase information of S_{11}. The total M-matrix of the whole structure can be expressed as the product of interface and layer matrices that describe the effects of the individual interfaces and layers of the entire stratified structure, taken in proper order, as:

$$M = \begin{bmatrix} M_{11} & M_{12} \\ M_{21} & M_{22} \end{bmatrix} = I_{01}L_1I_{12}L_2I_{23}L_3I_{34}L_4...I_{j\kappa}L_\kappa, \tag{10}$$

with:

$$J_{jk} = \frac{1}{t_{jk}}\begin{bmatrix} 1 & r_{jk} \\ r_{jk} & 1 \end{bmatrix}, \qquad r_{jk} = \frac{\left(\frac{K_{zj}}{\varepsilon_j} - \frac{K_{zk}}{\varepsilon_k}\right)}{\frac{K_{zj}}{\varepsilon_j} + \frac{K_{zk}}{\varepsilon_k}}, \tag{11}$$

where t_{jk} and r_{jk} are the Fresnel complex transmission and reflection coefficients, respectively. Also, the layer matrix (phase matrix) describing the propagation through layer j is defined as:

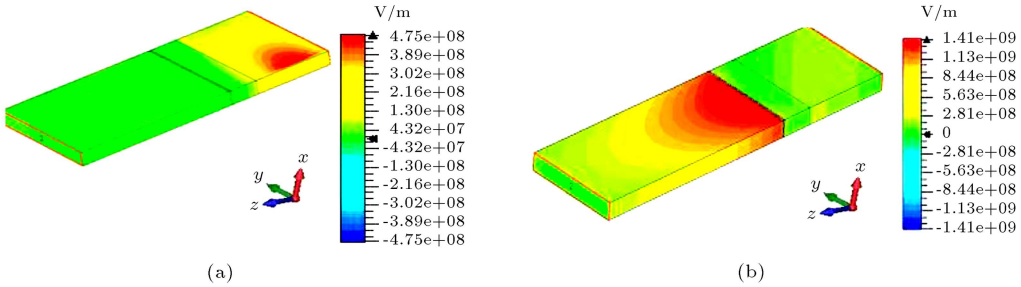
$$L_j = \begin{bmatrix} e^{ie^{id_zK_{zj}}} & 0 \\ 0 & e^{-id_zK_{zj}} \end{bmatrix}, \tag{12}$$

with:

$$K_{zj} = \sqrt{\left(\frac{\omega}{c}\right)^2\left[\varepsilon_j - \varepsilon_0\sin^2\theta\right]}, \tag{13}$$

where ε_j and d_j are the dielectric constant and the thickness of the jth layer, respectively.

For the detection of biomolecules, variations of S-parameters along with their phase characteristics resulting from biomolecular interactions with the sensing surface are studied in this paper. Thus, in order to investigate the sensor performance, we have studied the change of S-parameters for detection and sensitivity enhancement. In our proposed design, the resonances before and after binding occur at the incidence angles of 47.052 and 47.506 degrees, respectively. As shown in Figure 2, before the resonance with the angle of 45 degrees, the intensity plasmonic wave is reflected back whereas at the resonance condition with the angle of 47.506 degrees, the oscillation of plasmonic wave is maximum.



(a) (b)

Figure 2. Absolute electric field: (a) Before and (b) at resonance condition.

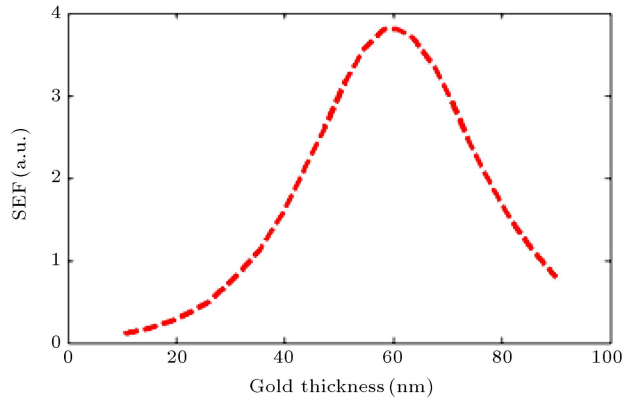


Figure 3. The peak values of SPR sensitivity as a function of the gold thickness changing from 10 to 90 nm.

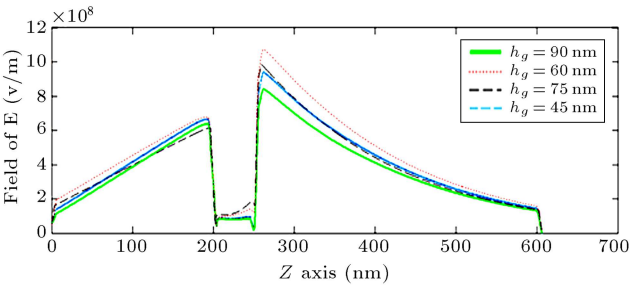


Figure 4. Vertical field along the z axis, E_z, when the SPR structure processes the gold layer with the thicknesses of 45, 60, 75, and 90 nm.

Figure 3 shows the peak values of sensitivity as a function of gold thickness in the range of 10 to 90 nm. It is clearly observed in this figure that the maximum sensitivity has been obtained as high as 3.82 at an optimal thickness of 60 nm. This is because the thicker gold layer leads to a resonance excitation of surface plasmons in a higher momentum.

The vertical field along the z axis (E_z) for the gold layer thicknesses of 45, 60, 75, and 90 nm is shown in Figure 4. According to this figure, as the gold layer thickness increases from 10 to 90 nm, the sensitivity increases at the initial stage with increase in grating thickness. However, when it exceeds 60 nm, the sensitivity begins to decrease (see [14] for more information).

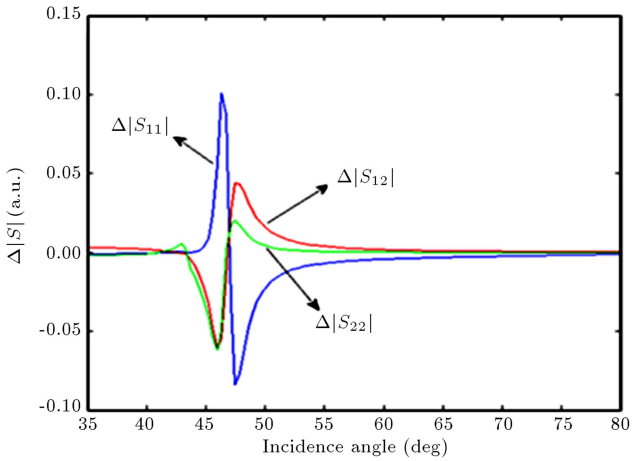


Figure 5. Changes of different S-parameters versus the incidence angle.

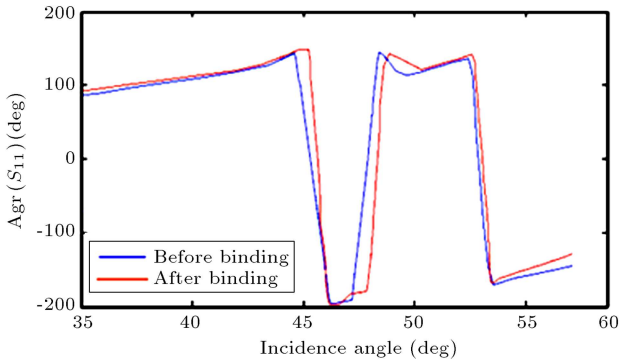


Figure 6. Phase variation of S_{11} (arg S_{11}) versus the incidence angle before and after the biomolecular interaction.

Figure 5 shows the changes of S-parameters versus the incidence angle for the proposed SPR biosensor. It is outlined that the maximum values of changes in S-parameters, $\Delta|S_{11}|$, $\Delta|S_{12}|$, $\Delta|S_{22}|$, and $\Delta|S_{21}|$, appear at their resonance angles and the greatest change is related to the S_{11} parameter.

In Figure 6, the phase plot of S_{11} versus the incidence angle before and after the biomolecular interaction is shown. It is observed that there is a rapid transition in the phase of S_{11} at the resonance angle of 47.052 degrees before binding.

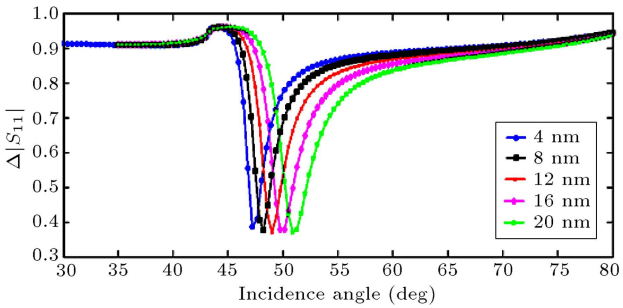


Figure 7. Various graphs demonstrating the effects due to the thickness of target biomolecules on the $|S_{11}|$ parameter.

Figure 7 shows the investigation of $|S_{11}|$ parameter changes versus the incident angle for different thicknesses of graphene layer. As can be observed in this figure, with increase in the thickness of graphene layer, the magnitude of S_{11} begins to decrease.

4. Conclusions

In this paper, we presented the design and analysis of a multilayer SPR biosensor using the FDFD method. The ssDNA was used as a receptor molecule, which was mobilized on the sensor surface to specifically adsorb its complementary counterpart (cDNA) immersed in a phosphate buffer solution. With employing the S-parameters-based detection, it was demonstrated that the proposed structure would improve sensitivity and speed of detection.

References

1. Choi, S.H., Kim, Y.L. and Byun, K.M. "Graphene-on-silver substrates for sensitive surface plasmon resonance imaging biosensors", *Optics Express*, **19**(2), pp. 458-466 (2011).
2. Jung, W.K., Kim, N.H. and Byun, K.M. "Numerical study on an application of subwavelength dielectric gratings for high-sensitivity plasmonic detection", *Applied Optics*, **51**(20), pp. 4722-4729 (2012).
3. Available at: https://www.cst.com/.
4. Lin, I.T. "Optical properties of graphene from the THz to the visible spectral region", Thesis of Master of Science, University of California, Los Angeles, UCLA (2012).
5. Wu, L., Chu, H.S., Koh, W.S. and Li, E.P. "Highly sensitive graphene biosensors based on surface plasmon resonance", *Optics Express*, **18**(14), pp. 14395-14400 (2010).
6. Mandenius, C.F., Chollet, A., Mecklenburg, M., Lundström, I. and Mosbach, K. "Optical surface methods for detection of nucleic acid binding", *Analytical Letters*, **22**(15), pp. 2961-2973 (1989).
7. Islam, M.S., Kouzani, A.Z., Dai, X.J., Schiretz, H.F., Lim, Y.C. and Michalski, W.P. "Sensitivity analysis of a sub-wavelength localized surface plasmon resonance graphene biosensor", *Complex Medical Engineering (CME)*, **2012**, *ICME International Conference on. IEEE* (2012).
8. Palik, E.D., *Handbook of Optical Constants of Solids*, **3**, Academic press (1998).
9. Taft, E.A. and Philipp, H.R. "Optical properties of graphite", *Physical Review*, **138**(1A), pp. A197-A202 (1965).
10. Neto, A.C., Guinea, F., Peres, N.M., Novoselov, K.S. and Geim, A.K. "The electronic properties of graphene", *Reviews of Modern Physics*, **81**(1), pp. 109-162 (2009).
11. Kim, K., Yoon, S.J. and Kim, D. "Nanowire-based enhancement of localized surface plasmon resonance for highly sensitive detection: A theoretical study", *Opt. Exp.*, **14**, pp. 12419-12431 (2006).
12. Homola, J. "Electromagnetic theory of surface plasmons", *Surface Plasmon Resonance Based Sensors*, **4**, pp. 3-44 (2006).
13. Islam, M.S. and Kouzani, A.Z. "A S-parameters-based detection method for a multilayer SPR biosensor", *Annual International Conference of the IEEE Engineering in Medicine and Biology Society, IEEE* (2012).
14. Majdi, M. and Fathi, D. "Improvement of the sensitivity of nano biosensors by multilayered structures", *24th Iranian Conference on Electrical Engineering (ICEE)* (2016).

Biographies

Mahdieh Majdi received the MSc degree in Optoelectronics from Tarbiat Modares University (TMU), Tehran, Iran, in 2015. Her research interests include graphene plasmonics, plasmonic waveguides, plasmonic gates, and switches.

Davood Fathi received the BSc degree in Electronic Engineering from Amirkabir University of Technology, Tehran, Iran, in 1990, and the MSc degree in Biomedical Engineering from Sharif University of Technology, Tehran, Iran, in 1994. After some years of working in industry, he worked toward the PhD degree between 2006-2009 in Nanoelectronics with the Nanoelectronic Center of Excellence, Thin Film and Photonics Research Laboratory, School of Electrical and Computer Engineering, University of Tehran, Tehran, Iran. Dr. Fathi joined the School of Electrical and Computer Engineering at Tarbiat Modares University (TMU), Tehran, Iran, in 2010, as a faculty member. His current research interests include nanoelectronics, nanophotonics and optoelectronics, solar cells, and quantum transport. He is also author or coauthor of more than 50 journal and conference papers in various fields of nanoelectronics and nanophotonics.