Glycosuria sensing based on nanometric plasmonic polaritons

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Abstract. Surface Plasmon waves exhibit an interesting characteristic of enclosing light above the diffraction limit. In this work surface plasmon sensor using metal-insulator-metal configuration is demonstrated for an application of Glycosuria detection. The sensing device possesses a slot waveguide and an elliptic cavity in a metallic substrate. On injection of TM polarized mode, surface Plasmon wave coupled with cavity causes the generation of resonating modes the shift in resonant frequency is utilized to detect glucose concentration in urine. The device's operation is investigated using the Finite Difference Time Domain (FDTD) method. The proposed sensor's sensitivity is evaluated to be 792.80nm/RIU, with a figure of merit of 113.25. Based on the analysis, it is remarked that the device is suitable for an on-chip application.

Keywords: Surface Plasmon, refractive index sensor, biosensor

1. Introduction

The high concentration of glucose causes serious complications in the human body such as kidney disorder, heart-related issues, blindness, etc. but for immediate energy supply, glucose is an important biomolecule [1,2]. While monitoring the concentration of blood glucose is normal, monitoring the concentration of urine glucose is also important to prevent chronic kidney disease [2]. So Glycosuria is the term used for the glucose content present in the urine[3,4].

For detection of biomolecules present in the body photonic biosensor present an intuitive device for the potential lab-on-chip applications. There are two types of Photonic biosensors: labeled and label-free[5]. Mostly labeled based sensor is used in which targeted biomolecules is labeled through a specific radioactive or fluorescent material. However it requires complex quantitative analysis and laborious process for accurate detection of biomolecules, also attachment of radioactive materials causes the alteration in the actual properties of the molecules that significantly deviates the output of the bio-sensing device[6,7]. The more enhanced method is label-free method that requires a specific bio-recognition element (BRE) layer for an attachment of the targeted analyte[8]. The event of attachment of the analyte to BRE causes changes in the material's physical properties, such as refractive index, and that modifies the overall optical characteristics of the device[7]. Recently, the devices work on the principle of the surface plasmon resonance (SPR) showing the great spark in development-of-nanometric-biosensors. These miniature biosensors have the capability of confining light, which is beyond the diffraction limit of the other optical devices[9]. The SPR based sensor have two configuration which are MIM (metal-insulator-metal) and IMI (insulator-metal-insulator) in both of these configuration the intensified electric field (Surface Plasmon Wave) confine in metal-insulator
interface. These surface plasmon waves are sensitive to change in refractive index of material under test [10–12]. Several researchers have demonstrated the use of SPR for bio sensing applications especially dependent on the blood components [9,12,13]. In this paper, a glucose-sensing system is investigated, which consists of a metal-insulator-metal (MIM) waveguide coupled with an inline elliptic cavity. Analysis of the device is performed using the FDTD method with a consideration of the perfectly matched boundary (PML) conditions to avoid any reflections from the outer boundaries of the region.

2. Modelling of Plasmonic Device

The proposed device structure is a result of etching performed on Silver metal to create a slot waveguide and an elliptic cavity, as shown in figure 1. The silver metal is placed on a Silicon Dioxide (SiO2) substrate. The typical dimension of the parameters present in the device is as follows: slot gap is represented by ‘g’ is of 50 nm, the major radius (R1) and minor radius (R2) of the elliptic cavity is 250 nm and 200 nm respectively, and coupling distance ‘d’ is of 20 nm. The disk cavity couples light in fraction, and hence act as a resonator in our device. The method are prescribed for the fabrication of these type of device structures is nano imprint lithography which uses focused ion beam milling technique [14].

![Figure 1. The basic structure of the Plasmonic device, $P_{in}$ and $P_{out}$ are input and output power respectively.](image)

The metal for a plasmonic device is of critical importance as it should have suitability in terms of the ease of fabrication, power losses, etc. [9]. The most profoundly used metals are Gold and Silver. In this work, Silver is considered due to the features that cause lower losses at the near-infrared region of the spectral region [15], and the frequency dependent material properties is characterized using the Drude model which is given as [9],

$$\varepsilon_{metal}(\omega) = \varepsilon_{\infty} - \frac{\omega_p^2}{\omega^2 + i\gamma\omega}$$

(1)

Where $\varepsilon_{metal}$ is the relative permittivity, the dielectric constant ($\varepsilon_{\infty}$) is 3.7, the plasma frequency is represented by $\omega_p = 9.1eV$, and $\gamma = 0.018eV$ is the free electrons collision-frequency and $\omega$ is the incident wave frequency. $P_{in}$ and $P_{out}$ represents the input and output power couple in the device, the transmission spectra (T) is defined by the expression $T = P_{in}/P_{out}$.

3. Computed Results

The FDTD approach is used to analyse the system, which is done with the help of the Opti FDTD simulation tool. The grid size in the simulation domain is of the size 3nm×3nm in X-Z direction, and
the time duration up to which simulations runs is set to be $\Delta t \leq \left( \sqrt{\Delta x^2 + \Delta z^2} \right) / c$ so that complete excited electric field decayed to zero \[16\]. For an initial investigation of the transmission spectra, the slot and elliptic cavity region of the device is assumed to be filled with the material having refractive index of 1 (air). When the device is excited through a TM polarized wave, the propagated SP mode travels along the metal-dielectric interface which allows the signal to evanescently couple to a cavity, the typical expression for the calculation of the resonant wavelength is $m \cdot \lambda_m = \left( 6 \cdot n_{ef} \cdot L \right) / \left( \frac{\omega}{v} \right)$, where $\theta$ is the phase shift observed by the propagating wave, ‘$m$’ is the order of mode, $L = 2\pi \sqrt{R_1^2 + R_2^2} / 2$ is the perimeter of the elliptic cavity, and the effective refractive index of the mode $n_{ef} = \beta_{SPP} / k_0$, which is found by considering the dispersion relation for the coupled TM mode which is expressed as[4,17],

$$\varepsilon_{metal} \cdot k \tanh \left( \frac{\omega \cdot k}{2} \right) + \varepsilon_{diet} = 0 \quad (2)$$

$$k_{dietmetal} = \sqrt{\beta_{SPP}^2 - \varepsilon_{dietmetal} \cdot k_0^2} \quad (3)$$

Where $\beta_{SPP}$ and $k_0$ are the propagation constant and the electric field wave vector respectively. The typical transmission spectrum is shown in figure 2. Two resonant peaks at wavelengths 579.6 nm and 804.84 nm are observed.

![Figure 2. Transmittance spectra of the device](image)

To utilize the structure for glucose sensing, the material under test is required to be filled in the device by using the capillary attraction [14], for an analysis purpose the variation in glucose concentration in urine is opted here in this work by the consideration of refractive index which is specified in table 1 [4].

**Table 1. Variation in refractive index for different glucose concentration in urine**

| Glucose Concentration in Urine | Refractive Index |
|-------------------------------|------------------|
| 0-15g/dl (normal)             | 1.335±0.0001     |
| 0.625g/dl                     | 1.336±0.0001     |
| 1.250g/dl                     | 1.337±0.0001     |
| 2.500g/dl                     | 1.338±0.0001     |
| 5.000g/dl                     | 1.341±0.0001     |
| 10.000g/dl                    | 1.347±0.0001     |

Figure 3 shows transmission spectra for variations in glucose content in urine, with a noticeable red change in spectra observed as the refractive index of the substance under examination is increased.
However, when plotting the resonant wavelength for variations in refractive index, figure 4 shows almost linear behaviour.

![Figure 3. Transmission-spectra of a device for different glucose concentration](image)

![Figure 4. Relationship between changes in resonant wavelength for a variation in refractive index](image)

The performance of plasmonic based photonic sensor is quantified through two characteristics, first is the sensitivity (\(S = \Delta \lambda / \Delta n\)) which represent the shift in resonant wavelength on account of the change in refractive index of material under test. Second is the Figure-of-Merit (\(FOM = S / FWHM\)) that is the ratio of S to 3-dB bandwidth (FWHM) of resonant peak. The sensitivity of the device is evaluated using figure 4 and is found out to be 792.8 nm/RIU and the FOM is 113.25 for 7 nm FWHM.

4. Conclusion
In this work, a glucose sensor based on Plasmonic Polaritons is reported. Based on the optimal parameters, the transmittance spectra show two resonating wavelengths at 579.6 nm and 804.84 nm. The simulation results represent that the sensing device possesses a linear relationship in terms of change in resonant wavelength on account of variation in change of refractive index of the glucose content in urine. The sensitivity of 792.8 nm/RIU is achieved with a figure of merit of 113.25. Hence this miniature sensing device is suitable to detect of glucose concentration in urine of the human body.
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