High sensitivity hollow core circular shaped PCF surface plasmonic biosensor employing silver coat: A numerical design and analysis with external sensing approach

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**ABSTRACT**

This article numerically offers and analyses a hollow core circular shaped photonic crystal fiber-based surface plasmon resonance (CH-PCF-SPR) biosensor. The biosensor sensitivity is analyzed with applying a mode solver built finite element method (FEM) incorporating multi-physics software “COMSOL”. A nano film of silver is coated as sensing layer on the external surface for easy sensing and more practical realization. The biosensor unveils the highest wavelength sensitivity valued 21000 nm/RIU and amplitude sensitivity valued 2456 RIU\textsuperscript{−1}, and corresponding wavelength resolution of 4.76 \times 10^{-6} RIU and amplitude resolution of 4.07 \times 10^{-6} RIU, respectively, using wavelength interrogation technique (WIT) and technique of amplitude interrogation (AIT), in detecting effective refractive index range 1.33 RIU and 1.42 RIU. Besides, the consequence of changing structural parameters like – pitch, diameter of air hole, various plasmonic metals and silver (Ag) layer thickness are also resulted in results section. The sensitivity of the offered sensor is analyzed to examine the means of spectral loss depth, resolution and amplitude sensitivity. In arrears to the modest scheme, highly sensitive and resolution nature, the offered design can be precisely applied in detection of bio molecular analytes

**Introduction**

The Instrumentations that are accompanying to detect the real time analyte concentrations are well defined as sensor. In opto-electronics based sensing purpose, surface plasmonic resonance (SPR) based refractive index bio-instrumentations are extremely recommended. The plasmonic sensor is exceedingly superior and has been widely applied to pharmaceutical findings, antigen–antibody collaboration, bio-imaging, food eminence measurement, environmental observation, and etc [1].

The united oscillation of surface electrons due to the existence of time varying EM field in-between metal/dielectric interface is well specified as SPR [2–4]. Plasmons are impartial explanations of Maxwell’s equations entailing in joint oscillations of charge, that be in motion comprehensively with a general frequency and propagation constant [5–7]. At the same time as majority plasmons have linked to wholesale systems, surface plasmons waves (SPW) which propagates by side of the metal/dielectric interface [8–10]. At what time the phase matching between optical wave and SPW is pleased, on the surface excelled optical wave gets joined with the SP modes of the metal/dielectric interface charitable birth to propagation oscillation along vertical direction and momentarily falling in longitudinal direction [11]. The circumstance of phase identical is originated to be aligned simply with the transverse magnetic (TM) polarization wave for the optical incident light [12].

Seeing as the straight optical signal has unable to bring sufficient...
energy that can stimulate SPPs along the expected interface, in earlier SPR techniques, Kretschmann configuration based on prism linking to metallic film was used as the SPR designs [13]. As a consequence of bulky size, unwarranted charge of cost of optical and mechanical components, remote sensing inability, irregularity of smallness, these conventional SPR instrumentations have not been acceptable.

To diminish these restrictions, Jorgenson in 1993 first reported a PCF-SPR based bio-instrumentation. The operating standard of PCF-SPR biosensor can be characterized by the mutual dealings between surface plasmon polariton (SPPs) electron and momentary field [13]. The specific wavelength at which the real part of the effective refractive index \( n_{eff} \) of real time analyte of fundamental core mode as well as SPP mode turns into analogous, is called resonance point. The impact of core-clad momentary field is the causes of the release of SPP electron from the metal/dielectric interface. Accordingly, the SPWs have been fashioned. In the application of SPR sensing, Au is favored as the greatest striking plasmonic material but it has low sensing range [14]. Therefore, gold can be replaced by another plasmonic metal named Silver (Ag) in [15] which is used for higher sensitivity [16].

To examine the sensitivity, it would be kept in mind that the design express a trouble-free manufacture process and extremely high performance characteristics. Lu studied in [14], SPR-PCF design based D-shaped biosensor which showed wavelength sensitivity of 3340 nm/RIU. As a result, surface polish excessively, creates manufacturing toughness. As a purpose of external detection in [17], Hasan and his team reported a sensor having maximum amplitude sensitivity of 371.5 RIU \(^{-1}\) and wavelength sensitivity of 4600 nm-RIU \(^{-1}\) [17]. However, here in [17] the amplitude sensitivity is not much high as compared the others.

Till date, a quantity of PCF-SPR biosensors have been studied for enlightening the sensitivity (S), Detection Accuracy (DA), Quality factor (QF) and vibrant range of detection etc. Most of the PCF-SPR biosensors are considered by two classes. Firstly, PCF-SPR sensors with internally fixed analyte thickness of 1.5 µm was introducing for the analysis [13] and dispersion relation are shown in a same Figure of [17]. The con

theoretical modeling

The cross-sectional schematic and graphical presentation of the offered hollow core circular shaped PCF-SPR biosensor is figurally described in Fig. 1. Here the sensor pattern has been modeled having five hexa-gonal shaped circles with total circular air hole (s). One air hole is omitted from the fourth circle to create a hollow core. The 4 air-holes (which are 2 taken from the third circle and 2 taken from the fifth circle) are set below for minimizing the air gap between the detecting dielectrics and core. For that reason, SPPs can surely be interacted to metallic surface simply. In order to create the circular ring, the side part in hexa-gonally photonic crystal air-holes have been vanished. In this modelling, the center-center distance between adjacent of two air holes is identified by pitch and represented by \( \Lambda \), the dia of small air hole (s) can be symbolized as \( d_0 \) and \( d \) indicates the dia of large air hole (s).

Here, firstly these variables are valued as, \( \Lambda = 3 \) µm, \( d_0 = 0.35 \Lambda \) and \( d = 0.82 \Lambda \). The silver layer with thickness; \( t_s = 30 \) nm is used as the plasmonic material which is put at planer surface of the device utilizing the chemical vapor deposition (CVD) method. As a, background material, we selected fused silica.

The famous Drude-Lorentz equation is practiced to calculate silver dielectric constants which are given in [17,20] as:

\[
\varepsilon_{Ag} = \varepsilon_{m} - \frac{\omega^2D}{\omega(\omega + j\omega L)} - \frac{\Delta \varepsilon \varepsilon_0 \Omega \Gamma}{(\omega^2 - \Omega^2L) + j\Omega^2L}
\]

where \( \varepsilon_{Ag}, \varepsilon_{m}, \omega_0, \omega \), and \( \gamma_0 \) represent silver permittivity, high frequency Ag permittivity, plasma frequency, angular frequency and decaying frequency, correspondingly. The specification of these variables is reported in literatures [17,20].

The Sellmier formula remains practiced to compute the refractive index of fused silica that was reported by M.B. Hossain et al., in [20].

\[
n^2(\lambda) = 1 + \frac{B_1\lambda^2}{\lambda^2 - C_1} + \frac{B_2\lambda^2}{\lambda^2 - C_2} + \frac{B_3\lambda^2}{\lambda^2 - C_3}
\]

Here, \( n \) is the effective RI of fused silica that is expressed in µm. The B1, B2, B3, C1, C2, C3 have the variables defined in literatures [20,21,22].

Theoretical results and performance analysis

A fixed analyte thickness of 1.5 µm was introducing for the analysis perseverance. The circular perfectly matched layer (CPL) a boundary condition is surrounded on thon the above of the sensing analyte. The triangular mesh size was kept considerably small as possible to ensure mapping the smaller air-hole correctly during simulation. The meshed computational domain has been covered by total of 19,426 triangular elements to present the biosensor where edge elements are1666 and vertex elements are 76. As well, the whole mesh area has been kept to 144.7 µm\(^2\).

The modal study is analyzed along XY plane as well as along the Z direction, the light is penetrated. The X-polarized mode shows in comparison better intensity of optical signal dissemination on the silver plane. Therefore, the momentary light signal can simply travel via the silver coating. Subsequently, the X components have considered as fundamental core for on the whole of computational simulation principle. The E and dispersion relation are shown in a same Figure of Fig. 2. At this point, refractive index of sensing analyte is 1.41 RIU. Here the coincide point between real refractive index in the elementary x-polarized mode and SPPs mode have been happened at a wavelength of 0.85 µm. And this wavelength is called resonating wavelength. In this point, a deep sharp reasoning peak has been observed that denotes the energy transfer is maximum from PCF core to SPS.

Sensor performance study on the effect of analyte Variation.

At first in this paper, we consider the changing characteristics of confinement loss and amplitude sensitivity which both play a significant role in sensor performance due to the variation of effective RI. The confinement loss can be obtained by means of the following formulation [23].

\[
\alpha(\text{dB/cm}) = 8.686 \times k_0 \text{Im}(n_{eff}) \times 10^4
\]

where \( \text{Im}(n_{eff}) \) is the imaginary core mode RI, \( k_0 = 2\pi/\lambda \) is the wave number, \( \lambda \) expresses working wavelength. The change of sensing medium effective RI deeply impact on the effective refractive index real
An unidentified sensing dielectrics can be simply sensed due to the smaller changing of refractive index by observing the shift of resonating peak to the greater wavelength.

From Fig. 3, it is being seen that owing to the change of $n_a$ from 1.33RIU to 1.34RIU resonance wavelength is shifted from 0.52 μm to 0.54 μm, from 1.34RIU to 1.35 RIU resonance wavelength is shifted from 0.54 μm to 0.57 μm, from 1.36RIU to 1.370 RIU resonance wavelength is changed from 0.57 μm to 0.60 μm, from 1.37RIU to 1.38 RIU resonance wavelength is changed from 0.60 μm to 0.64 μm, from 1.38RIU to 1.39 RIU resonance wavelength is changed from 0.64 μm to 0.69 μm, from 1.39 RIU to 1.40 RIU resonance wavelength is changed from 0.69 μm to 0.76 μm, from 1.40 RIU to 1.41 RIU resonance wavelength is changed from 0.76 μm to 0.86 μm and from 1.41 RIU to 1.42 RIU resonance wavelength is changed from 0.86 μm to 1.07 μm, respectively. From Fig. 3(a), least confinement loss 2.5 dB/cm has been found at the wavelength of 0.5 μm in RI of a real time analyte of 1.33 RIU, and highest peak loss value of 161.73 dB/cm was observed at the wavelength of 1.07 μm for the analyte RI of 1.42RIU. The wavelength sensitivity of the offered biosensor is obtained by using the WIT and defined as following [15,17,20]:

$$\Delta \lambda_{\text{peak}} / \Delta n_a = S_1(\text{nm}/\text{RIU}) = \Delta \lambda_{\text{peak}} / \Delta n_a$$

Here $\Delta \lambda_{\text{peak}}$ symbolizes wavelength peak difference and $\Delta n_a$ symbolizes RI change of analyte. The offered design has been shown a variance shift sandwiched between peak wavelength and RI. The variation of peak wavelength ($\Delta \lambda$ peak) has been seen of 20, 20, 30, 40, 50, 70, 90, and 210 nm with respect to the RI variations ($\Delta n_a$) from 1.33 to 1.34, 1.34 to 1.35, 1.35 to 1.36, 1.36 to 1.37, 1.37 to 1.38, 1.38 to 1.39, 1.39 to 1.40, 1.40 to 1.41, and 1.41 RIU to 1.42 RIU, correspondingly. So, the numerical wavelength sensitivities stand computed of 2000, 2000, 3000, 3000, 4000, 4000, 5000, 7000, 9000, and 21000 nm/RIU, respectively, by using Eq. (4). Another important factor is “resolution” of sensor that expresses how sensor can sense a minimum variation in RI of a real time analyte. The Eq. (5) computes the resolution studied in literature [20,22]:

$$R(\text{RIU}) = \Delta n_a \times \Delta \lambda_{\text{min}} / \Delta \lambda_{\text{peak}}$$

Fig. 1. (a). 2D Stacked of the offered design along the X-Y plane. (b) 2D cross sectional presentation of the hollow core circular shaped PCF-SPR biosensor with $\Lambda = 3 \mu m$, $d_s = 0.35\Lambda$, $d = 0.82\Lambda$ and $t_s = 30$ nm.

Fig. 2. (a). Dispersion relation between fundamental core mode and SPP mode. The electric field distribution between (b) Core guided mode and (c) Plasmonic or SPP mode at wavelength of 0.77 μm and for real time analyte RI ($n_a$) = 1.41 RIU, $t_s = 30$ nm and $\Lambda = 3 \mu m$, $d = 0.35\Lambda$, $r = 0.82\Lambda$.

Fig. 3. (a) The loss spectrum as well as (b) amplitude sensitivity with respect to RI of real time analyte variation from 1.33 RIU to 1.42 RIU having $\Lambda = 3 \mu m$, $d = 0.35\Lambda$, $r = 0.82\Lambda$, and $\Lambda = 30$ nm.
Assuming $\Delta n = 0.01$, $\Delta \lambda_{\text{min}} = 0.1$ nm, as well as $\Delta \lambda_{\text{peak}} = 210$ nm, here the maximum wavelength resolution of the CH-PCF-SPR biosensor is obtained about $4.76 \times 10^{-6}$RIU by using equation (5). Therefore, the offered bioinstrumentation capable exactly detect sensing medium dielectrics refractive index deviation in $10^{-6}$order. In order to know the sensitivity, the wavelength interrogation technique utilizes FWFM management that's ensures sensing procedure difficult but another method, for example, difference in amplitude can reduce this limitations by calculating the sensitivity at a specific wavelength. The below formulation explains the amplitude sensitivity [15,20]:

$$S_a(\text{RIU}^{-1}) = -\frac{1}{\alpha(\lambda, n_a)} \frac{\partial (\alpha, n_a)}{\partial n_a}$$

where $\alpha(\lambda, n_a)$ denotes the propagation loss and $\partial (\alpha, n_a)$ denotes the difference of loss. In Fig. 3(b), the amplitude sensitivity curve has been shown for different RIs of analyte. An utmost amplitude sensitivity of 2456 RIU$^{-1}$is resulted at 1.07 $\mu$m SPR wavelength having an $n_a$ of 1.41 RIU. In addition, the amplitude sensitivity about 147, 188, 251, 323, 434, 588, 815 and 1277 RIU$^{-1}$ have been reported for the sensing di-electrics refractive index of 1.45 RIU and 1.46 RIU, respectively. Now, letting we consider 1% variance in the transmitting intensity, an utmost device amplitude resolution of $4.76 \times 10^{-6}$ RIU is measured for the described sensor. The performance of the described design has the superior value comparable with the previously proposed other works [15–17].

Sensor performance study on the effect of silver layer variation

Silver layer has a noteworthy effect on sensor performance. Secondly, in this paper we have shown how the sensor layer thickness and loss spectrum can be changed with the silver layer thickness. Here, the silver layer thickness ($t_s$) stays tweaked from 30 to 50 nm and corresponding confinement loss spectra for real time detection of analyte RI of 1.39RIU and 1.40 RIU have been specified in Fig. 4(a).

We noticed that by increasing silver layer thickness causes increased of loss depth. Here, we noticed that the rise of silver coating thickness resulting in a changing of peak loss to greater wavelength. The amplitude sensitivity of 815, 500, and 294 RIU$^{-1}$ have been achieved for the $t_s$ of 30 nm, 40 nm, and 50 nm, accordingly, which is described in Fig. 4(b).

Sensor performance study on the effect of pitch variation

In third, pitch parameter ($\Lambda$) is taken into consideration on the loss spectra as well as amplitude sensitivity. The effect of pitch is seen in Fig. 5(a) and Fig. 5(b). It is shown in Fig. 5 (a) the loss spectrum is knowingly decreased with respect to the increase of pitch parameter because it increases the differences of effective RI between core-cladding. It is shown in Fig. 5, the raise of pitch parameter from 2.9 $\mu$m to 3.2 $\mu$m causes the maximum loss peak shifted higher wavelength from 7.76 dB/cm to 10.18 dB/cm and minimum loss peak from 5.22 dB/cm$^{-1}$ to 7.10 dB/cm$^{-1}$ having a change of $n_a$ from 1.380 to 1.39 RIU ($\Delta n_a = 0.01$ RIU). At the same configuration, maximum achievable amplitude sensitivity is reported to 591 dB/cm with $\Lambda = 3$ $\mu$m as depicted in Fig. 5 (b).

It stays clearly seen from Fig. 5(b), some little bit deviation of pitch parameter from 3 $\mu$m, the value of amplitude sensitivity is equitably dropped. As stated by the illustration of Fig. 5 (b), amplitude sensitivities are fallen to 537 dB/cm, 588 dB/cm, and 585 dB/cm, respectively, having a pitch value increased from 2.9 $\mu$m to 3.2 $\mu$m. So, the chosen optimum pitch parameter in this study is 3 $\mu$m.

Sensor performance study on the effect of air hole diameter variation

As a forth discussion parameter, here, the impact of air-holes diameter on sensor performance can be described by utilizing the change of diameter $d$ and $d_s$ as given in Fig. 6. It has been observed from Fig. 6 (a) that if the diameter of $d$ is increased, then the loss depth is also gradually increased. Results show that minimum loss peak of 5.33 7.54 dB/cm and 7.54 dB/cm have been reached at $d = 0.77 \Lambda$ having a change of $n_a$ from 1.390 to 1.4 RIU. Then again, maximum loss peak of 15.63 dB/cm$^{-1}$ and 20.55 dB/cm$^{-1}$ have been reached at $d = 0.87 \Lambda$ having a change of $n_a$ from 1.39 to 1.4 RIU. As seen in Fig. 6 (a) and (b), the red shade denotes the outcome by reason of the change of large dia ($d$). Moreover, the equivalent amplitude sensitivities are achieved to 703 dB/cm, 803 dB/cm, 815 dB/cm, and 844 dB/cm having a diameter of $d$ equals 0.77$\Lambda$, 0.80$\Lambda$, 0.82$\Lambda$.
and 0.87 λ as revealed in Fig. 6(b). Again, in Fig. 6(c) that shows the loss peak due to the change of small air holes diameter (ds) and in Fig. 6(d) that represents the analogous amplitude sensitivity. Here, uppermost sensitivity is attained at diameter (d) equal to 0.37 λ.

As revealed in Fig. 6(c) and (d), the red shift denotes the outcome due to the change of small air hole diameter ds. The loss peak is found to 7.8 dB/cm, 6.75 dB/cm, 6.17 dB/cm, and 4.97 dB/cm, respectively, having a small air hole diameter (ds) of 0.30 λ, 0.35 λ, 0.37 λ, and 0.40 λ with an n_a of 1.38 RIU. And loss depth is found to 9.91 dB/cm, 9.11 dB/cm, 8.68 dB/cm, and 6.22 dB/cm, respectively having a small air hole diameter (ds) of 0.30 λ, 0.35 λ, 0.37 λ, and 0.40 λ with an n_a of 1.39 RIU. Moreover, amplitude sensitivities are attained to 586, 588, 713, and 569 RIU⁻¹ with a small air hole diameter value of 0.30 λ, 0.35 λ, 0.37 λ, and 0.40 λ, respectively, with an analyte RI of 1.38 RIU. In spite of the utmost sensitivity is attained at the small air hole diameter (ds) of 0.37 λ, but for overall improved sensing performances, we carefully chosen small air hole diameter (ds) of 0.35 λ.

**Polynomial fitting**

A suitable sensor shows high linear characteristics. As a fifth factor in this paper, Fig. 7 presents the polynomial fitting response according to quick real time detection of analyte variation from 1.33 to 1.42 RIU. The polynomial fitting curve of offered design demonstrates R² is 0.7855 that's denotes admirable linearity.

Since its outstanding linear fitting belongings, the offered PCF-SPR design will be a probable applicant in medical industry, environment monitoring for quick real time detection of analyte.

**Performance comparison study**

The proposed circular shaped hollow core PCF-SPR design can easily be physically employed by utilizing typical "stack-and-draw" manufacture process described in [24–28].

At the end, we prepared Table 1 which shows a comparative study of performances among different plasmonic metal of the raised CH-PCF-SPR. The establishment of Table 2, has made with considering amplitude sensitivity, wavelength sensitivity, amplitude resolution and wavelength resolution. In Table 2, it reveals that our elevated CH-PCF-SPR sensor has attained the numerical amplitude and wavelength sensitivity of 2456 RIU⁻¹ and 21000 nm/RIU, respectively, which are an intensively large value than the traditional PCF sensor and that admits the

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**Table 1**

<table>
<thead>
<tr>
<th>Plasmonic material</th>
<th>Sensing ranging (RIU)</th>
<th>Wavelength sensitivity (nm/RIU)</th>
<th>Amplitude sensitivity(RIU⁻¹)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gold</td>
<td>1.33–1.39</td>
<td>5000</td>
<td>699</td>
</tr>
<tr>
<td>Copper</td>
<td>1.34–1.40</td>
<td>9000</td>
<td>1144</td>
</tr>
<tr>
<td>Silver</td>
<td>1.33–1.42</td>
<td>21,000</td>
<td>2456</td>
</tr>
</tbody>
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highest value in the literature. The larger amplitude sensitivity sensor will be very useful for remote biological, biomedical and biochemical analytes detection.

**Conclusion**

A simple circular hollow core PCF-SPR biosensor is offered and analyzed in this proper. For numerical and theoretical analysis, FEM has been used. The sensor has high sensitivity with comparatively low confinement loss for a long detection ranging 1.33 RIU to 1.42 RIU. Utilizing WIT and AIT, the obtainable utmost amplitude sensitivity of 2456 RIU$^{-1}$ and sensitivity of wavelength 21000 nm/RIU, respectively, and wavelength resolution of 4.76 × 10^{-6}RIU and resolution of amplitude of 4.07 × 10^{-6} RIU. So, we can use this sensor in detecting unknown analyte such as bio molecular elements various types of lipid, proteins and carbohydrates and sensing applications.

**Declaration of Competing Interest**

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.

**References**