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Performance Improvement of an Au and TiO_2 Coated D-shaped Photonic Crystal Fiber (PCF) Based SPR Sensor for Cancer Detection

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Abstract. Surface plasmon resonance (SPR) has been an emerging tool for the recognition of cancer in a label-free manner. However, conventional SPR setups are based on prism coupling, which makes them bulky, hindering their capability to be integrated into compact and portable devices. To overcome these issues, the proposed approach introduces a photonic crystal fiber (PCF) structure with a micro-channel D-shaped, integrated with a gold (Au) coating as the plasmonic substance and titanium dioxide (TiO_2) as a dielectric substance. It minimizes the distance of the plasmonic film from the fiber core, thus elevating the sensor performance. In addition, the introduction of TiO_2 over the plasmonic film alters the plasmon-analyte interaction, thus increasing the sensitivity. The numerical simulation for the designed sensor is done in COMSOL Multiphysics software through the finite element method (FEM). After optimization, the sensor is numerically evaluated for the identification of six various carcinoma cell types for refractive indices between 1.360 to 1.401. The ultimate achieved amplitude and spectral sensitivities are 717.29 RIU^{-1} and 7142.86 nm/RIU, respectively, recorded for MCF-7 and MDA-MB-231 cells. Additionally, the optimized sensor can distinguish a diverse array of biomolecules across the RI span of 1.330 to 1.420, attaining a peak spectral sensitivity of 20,000 nm/RIU for y-polarized light. Moving further, the sensor also delineated excellent figure of merit, resolution, and amplitude sensitivity values of 366 RIU^{-1}, 5×10^{-6} RIU, and 1178 RIU^{-1}, respectively, within the mentioned RI domain. The measured outcomes show that the designed sensor could be a potential option for the precise identification of various cancer cells.

Keywords. Cancer, finite element method (FEM), Photonic crystal fiber (PCF), refractive index (RI), surface plasmon resonance (SPR)

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1. Introduction

A total number of 618,120 cancer fatalities and 2,041,910 newly discovered carcinoma cases are estimated to occur in the USA by the end of 2025 [1], [2], [3], [4]. The global death toll projection is going up to 16.9 million per year by the year 2045. A significant portion of cancers exhibit no symptoms until later stages, demonstrating the disease's complex and challenging nature, with 70 % of cancer fatalities in resource-limited areas resulting from restricted access to early screening and treatment [5]. Despite significant advances in conventional cancer detection methods, such as imaging modalities like mammography, MR imaging, computed tomography (CT), positron emission tomography (PET), and biopsies, as well as blood and tissue-based diagnostic evaluation tools like polymerase chain reaction (PCR) and enzyme-linked immunosorbent assay (ELISA), these mechanisms still endure major drawbacks. These include low specificity, high costs, radiation exposure risk, time-intensive procedures, and dependency on advanced laboratory setup, among others [6]. While AI and machine learning have contributed to improved scan accuracy, these technologies still do not provide the kind of non-invasive, real-time, and universal cancer detection needed [7]. Taking into consideration the significant constraints that the traditional methods present, it is logical to say that the healthcare system needs quick, inexpensive, and non-invasive diagnostic alternatives. A biosensor could be a probable solution for overcoming the aforementioned limitations due to sensitive, selective, and time-efficient analysis of biological targets in a user-friendly manner. Present-day biosensors use enzymes and antibodies as biorecognition elements in their detection of cancer biomarkers [8], [9]. However, the use of enzymes and antibodies results in degradation over time, and subsequently, the sensor performance is affected.

In this context, optical biosensors can hold significant potential in cancer diagnosis owing to their real-time and continuous detection in a highly responsive, selective, and affordable, and non-invasive manner. Different optical biosensors also have dependence on labeled techniques, which incorporate electrochemically active probes, fluorescent labeling, and chemiluminescence. Nevertheless, this method can be quite expensive, require a lot of time, and may also disrupt the receptor-analyte interactions [10]. To avoid this problem, label-free optical sensors are strongly recommended. Surface plasmon resonance (SPR) biosensors that allow the identification of target substances at very low concentrations without the need for labeling are the most precise optical

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methods [11], [12]. The SPR serves as a powerful tool for point-of-care testing of oncogenic biomarkers in a label-free manner with high specificity, sensitivity, fast sensing, and high repeatability, which can also characterize distinct types of carcinomas. The principle behind the SPR biosensor is a shift in the refractive index (RI) in the ambient environment being measured as it relates to the binding of molecules at the surface of the metal. A heterostructured SPR biosensor composed of graphene and MXene ($Ti_3C_2T_X$), along with a BK7 prism, gold (Au), and aluminum oxide (Al_2O_3) for the rapid characterization of a well-known cancer-associated marker, carcinoembryonic antigen, was proposed by Khodaie and Heidarzadeh [13]. The findings presented a figure of merit (FOM) of 17.52 RIU^{-1} and a comparatively enhanced sensitivity of 163.63 deg/RIU. The use of ZnO and TMDCs layers in a SPR biosensor by Janze et al. [11] allowed for the identification of distinct carcinoma cells. The proposed biosensor demonstrated the figure of merit and the highest sensitivity values of 124.86 RIU^{-1} and 342.14 deg/RIU in terms of detecting blood cancer (Jurkat) cells from healthy cells. Another recent study proposed a $BaTiO_3$ and black phosphorus-based SPR biosensor for the identification of different malignancy biomarkers, where a maximum sensitivity of 417.86 deg/RIU was achieved for breast cancer MCF-7 cells (mammary carcinoma type II) [5]. On the other hand, the regular prism-coupled SPR setups, which are the basis for sensing technologies, have some drawbacks, such as their large size, high price, and the fact that they are difficult to incorporate into small, portable devices.

In this context, the fiber optic SPR sensors, particularly those utilizing Photonic Crystal Fiber (PCF), so to speak, are the ones bringing the solution to the problems because they are providing miniaturized, flexible, and robust platforms that are, at the same time compatible with in vivo applications and point-of-care diagnostics [14]. These enticing features make PCF-based SPR sensors desirable for various biosensing applications, including the detection of blood components [15], heavy metals [16], cancer cells [17], and various other diseases [18]. Additionally, the use of AI for geometrical optimization has further enhanced the potential of PCF-based SPR sensors for biomedical applications [19]. Among various fiber geometries, including circular, D-shaped, slotted, and bowl-shaped, the D-shaped structure can improve sensor performance by incorporating microchannels [20], [21]. Numerous researchers have reported a wide variety of D-shaped structures for detecting various analytes. For instance, Divya *et al*. [22] suggested a dual-channel D-shaped PCF-based plasmonic sensor that is able to sense two different

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analytes simultaneously. The sensor went through an optimization process and then reached the peak amplitude and wavelength sensitivities of 216 RIU^{-1} and 10000 nm/RIU, respectively. Another research work reported by Zhou *et al*. [23] was on a D-shaped PCF-based SPR sensor whose core was an open-ring channel sandwiched between a gold deposition to trigger the plasmonic modes. They showed that analyte placement significantly affects sensitivity. Specifically, when the target analyte was positioned across the outer ring structure, the sensor exhibited an RI finding that lies within the span of 1.2 to 1.34, achieving a peak sensitivity of 2000 nm/RIU. In contrast, when the analyte was placed only on the open channel, the measurable RI span extended from 1.33 to 1.46, and the sensor achieved a peak sensitivity of 5600 nm/RIU. To further improve the performance, Oudenani and Sonne proposed a D-Shaped SPR-Based PCF biosensor, which obtained an optimum amplitude and wavelength sensitivity values of 1623.6 $RIU^{-1}$and 12,300 nm/RIU, respectively [24]. On the other hand, Azadi *et al*. sandwiched a titanium dioxide (TiO_2) dielectric material between the gold (Au) and the silica for light trapping. The structure they suggested achieved an optimized wavelength sensitivity (WS) of 14,000 nm/RIU and amplitude sensitivity (AS) of 610 RIU^{-1}. Moving further, Azadi *et al*. [25] conducted a study wherein a TiO_2 dielectric coating was incorporated between a gold plasmonic layer and silica. This structure achieved an optimized AS of 610 $RIU^{-1}$and WS of 14,000 nm/RIU. Moreover, the D-shaped design is efficacious in terms of production as it does not require a complicated setup, as in the case of circular or dual-core PCF, besides being more durable with the Au or TiO_2 coatings, and allowing dual-polarization for further deterring of non-target interfering molecules with [13] increased specificity.

Such facts have prompted this study to go for a D-shaped sensor, which facilitates enhanced performance in distinguishing between different cancer cells and their normal counterparts. The FEM is used to develop and investigate the designed sensor. The study critically examined how the orientation of air spaces, their diameter, and the height of the plasmonic film and overlayers affected the overall effectiveness of the sensor. The influence of the open channel on sensor efficacy is also evaluated. After a thorough study and evaluation, it is concluded that the adjusted sensor structure can identify six different cancer biomarkers from the corresponding healthy cells with maximum amplitude and spectral sensitivities of 717.29 $RIU^{-1}$and 7142.86 nm/RIU, respectively, for MCF-7 and MDA-MB-231 cells. In addition, the optimized sensor exhibited a

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maximum sensitivity, figure of merit, and resolution values of 20,000 nm/RIU, 1178 RIU^{-1}, 366 RIU^{-1}, and 5×10^{-6} RIU, respectively, within a wide RI span from 1.330 to 1.420 for y-polarized light. This depicts the sensor's effectiveness in detecting a broad spectrum of bioanalytes with enhanced performance, illustrating its potential to be a strong tool for medical diagnostics.

2. Design of PCF Structure and Theoretical Model

2.1. Overview of the PCF framework

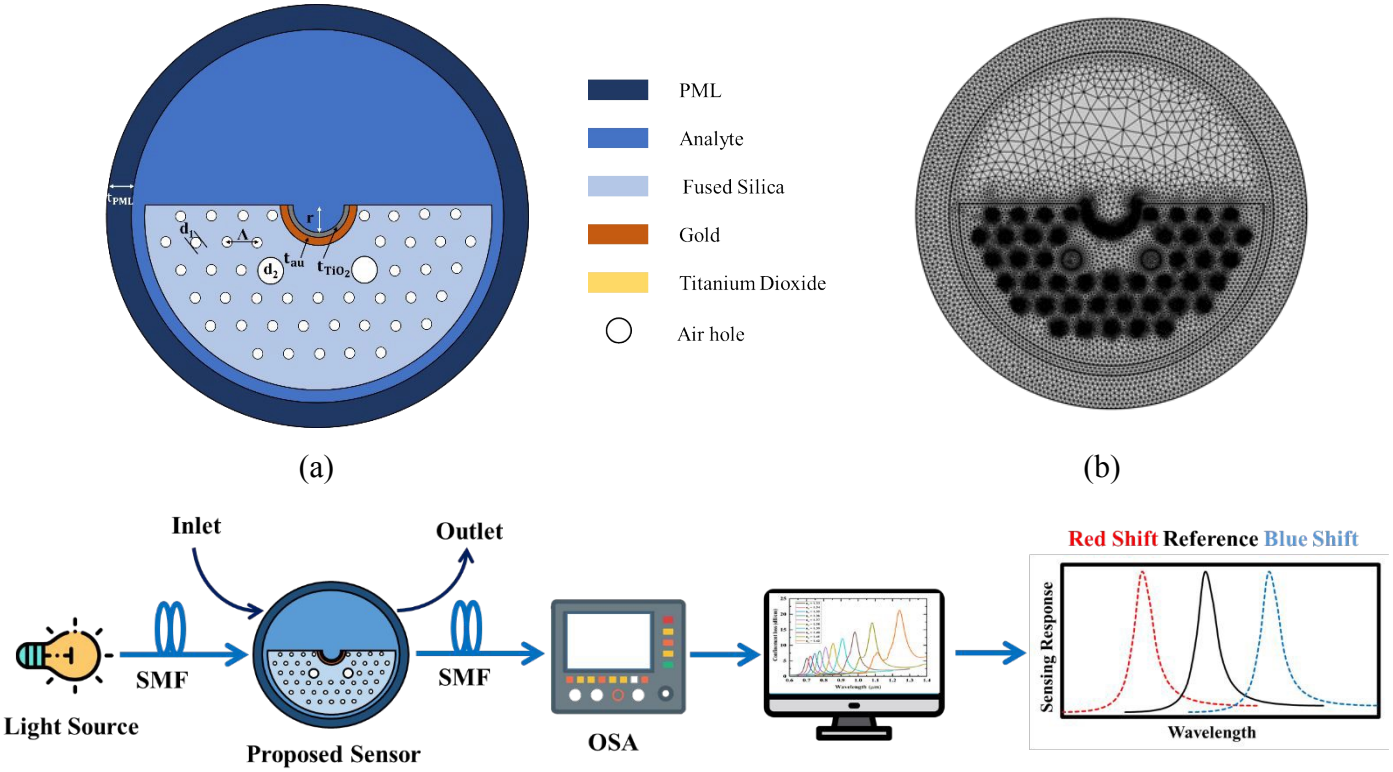

Fig. 1. Cross-sectional view of the (a) designed microchannel-type D-shaped sensor, (b) finite element mesh, and (c) experimental setup of the proposed sensor.

This study describes the design and numerical assessment of a SPR sensor based on PCF, utilizing the full-vector FEM in COMSOL Multiphysics 6.1. Fig. 1(a-b) depicts the schematics of the designed structure and finite element mesh. Simulations were conducted using a frequency-domain eigenmode solver with a non-uniform triangular mesh. A non-uniform triangular mesh with local refinement at the Au and Au-TiO_2 interface (192,032 domain, 6,136 boundary elements)

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ensured convergence. Electromagnetic boundary conditions included PML at domain edges, and continuity at material interfaces was implemented. Effective index and confinement loss were extracted from Im(neff) using default solver tolerances. The sensor comprises three main components: a silica (SiO_2) substrate featuring circular air spaces organized in a hexagonal lattice, a gold (Au) plasmonic layer, and an analyte channel. The cladding region comprises two separate types of air holes, labeled d_1 and d_2, meticulously designed for superior optical confinement and improved sensing efficacy. The midpoint-to-midpoint gap between neighboring air holes is labeled as pitch, which determines the modal properties and the guided mode's effective refractive index. A thin coating of TiO_2 is embedded between the Au layer and the analyte with a height of t_{TiO_2}. After conducting a series of optimization experiments, the structural parameters have been established as pitch = 4 μm, r = 8 μm, d_1 = 1.1 μm, d_2 = 3 μm, t_{au}= 55 nm, and t_{TiO_2}= 09 nm. t_{au} indicates the height of the Au. The D-shaped arrangement is attained by polishing the fiber surface near the core region, which minimizes the separation between the guided core mode and the Au plasmonic layer. This reduced distance enhances the penetration of the evanescent electric field into the plasmonic coating interface, increasing the modal overlap and improving phase matching with the SPP mode, thereby strengthening SPR excitation and excitation efficiency. The external section of the structure includes an analyte open microchannel with diameter r and a perfectly matched layer (PML) with a thickness of $t_{PML} = 5$ μm, to mitigate undesirable boundary reflections.

Fig. 1(c) explains a simplified representation of the experimental setting for the sensing approach, including an optical tunable source (OTS) and an optical spectrum analyzer (OSA). The designed sensor is interconnected via a single-mode fiber (SMF). The analyte is placed on the external surface of the sensor, which can be regulated using a pump that passes through the IN/OUT channel. The interaction between the unknown analyte and the sensor surface produces a resonance wavelength shift toward the red or a blue shift, as depicted in Fig. 1 (c). The OSA can monitor this phenomenon. The unidentified analyte can be determined by investing the confinement loss on a computer.

2.2. Material selection and theoretical basis

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Gold (Au) is an effective plasmonic substance due to its remarkable chemical stability, biocompatibility, and extensive plasmon resonance bandwidth, rendering it highly appropriate for sensing applications. The relative dielectric properties of Au are delineated by the Drude-Lorentz model [26].

$$\epsilon_{Au}(\omega) = \epsilon_{\infty} - \frac{{\omega_D}^2}{\omega(\omega + i\gamma_D)} - \frac{\nabla\epsilon \cdot {\Omega_L}^2}{(\omega^2 - {\Omega_L}^2) + i\Gamma_L\omega} \quad (1)$$

Where ϵ_{Au}, ϵ_{∞}, ω_D , γ_D, Γ_L, $\nabla\epsilon$, and Ω_L are the relative dielectric properties of gold, the high-frequency dielectric property, the plasmon frequency, the damping coefficient, the frequency band width of the Lorentz oscillators, the weighted parameter, and the oscillator strength, respectively, with their values specified in Table 1.

Table 1. Drude–Lorentz model parameters.

ϵ_{∞}	$\omega_D/2\pi$ (THz)	$\gamma_D/2\pi$ (THz)	$\Gamma_L/2\pi$ (THz)	$\Omega_L/2\pi$ (THz)	$\nabla\epsilon$
5.96730	2113.60	15.920	104.86	650.07	1.09

A thin TiO2 layer is applied as an optical tuning material on the gold coating to modify the local dielectric conditions at the metal-dielectric interface. Its high refractive index enhances electromagnetic field confinement and improves phase matching with the SPP mode, leading to stronger plasmon-analyte interaction and improved refractive index sensitivity. The subsequent equation determines the relation between the wavelength of TiO_2 and its refractive index [27].

$$n_{TiO_2}^2 = 5.913 + \frac{2.441 \times 10^7}{(Wl^2 - 0.803 \times 10^7)} \quad (2)$$

In this context, Wl denotes the operational wavelength in angstroms, and the titanium dioxide's RI is n_{TiO_2}.

Fused silica serves as the foundational substance of the structure in this research. The RI of Fused silica changes with the wavelength of light, which causes material dispersion when the fiber is in operation. Therefore, it is fundamental to perform numerical simulations to assess how material dispersion affects the PCF's transmission properties. The Sellmeier equation can determine the RI of fused silica [28].

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$$n^2(Wl) = 1 + \frac{E_1 Wl^2}{Wl^2 - {F_1}^2} + \frac{E_2 Wl^2}{Wl^2 - {F_2}^2} + \frac{E_3 Wl^2}{Wl^2 - {F_3}^2} \quad (3)$$

Where the RI is denoted as *n* and *Wl* is the wavelength of fused silica, measured in microns. The dispersion parameters of the Sellmeier equation are given in Table 2.

Table 2. Sellmeier dispersion parameters.

E_1	E_2	E_3	F_1	F_2	F_3
0.6961663	0.4079426	0.8974794	0.0684043	0.1162414	9.896161

The assessment of sensing performance in PCF-based SPR sensors is significantly dependent on the analysis of confinement loss. The subsequent formula can be used to determine the confinement loss [29].

$$\alpha(\frac{dB}{cm}) = 8.686 \times 10^4 \times \left(\frac{2\pi}{Wl}\right) \times Im(neff) \quad (4)$$

Where *Im(neff)* and *Wl* denote the fictitious segment of the effective index of the fundamental mode, and the incident wavelength, respectively.

The wavelength sensitivity (WS) analysis is a key parameter to assess the sensor's performance. The spectral interrogation approach evaluates sensitivity by observing the shift in resonance wavelength relative to the analyte's RI, mathematically represented as follows [30].

$$WS = \frac{\Delta Wl_{peak}}{\Delta n_a} \left(\frac{nm}{RIU}\right) \quad (5)$$

The shift in resonance wavelength, which occurs between two successive peaks, is represented by ΔWl_{peak} and the difference between the two RIs is represented by Δn_a.

The sensor performance can be evaluated by another significant measurement called amplitude sensitivity (AS). AS depends on variations in oscillatory strength and recognizes the analyte by assessing the changes in confinement loss. The formula for measuring AS is presented as follows [31]:

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$$AS(Wl) = -\frac{1}{\alpha(Wl,n_a)} \times \frac{\partial\alpha(Wl,n_a)}{\partial n_a} \ (RIU^{-1}) \quad (6)$$

The variation in confinement loss between two analytes is $\partial\alpha(Wl,n_a)$, and confinement loss for the analyte used for the detection is $\alpha(Wl,n_a)$. Within this context, ∂n_a denotes the change between the RIs of analytes that are adjacent.

Furthermore, the sensor's figure of merit (FoM) is evaluated to assess the sensing efficacy. The FoM is a quantitative indicator that tells the effectiveness of a sensor, with higher values indicating a higher degree of precision for identifying the RI of a solution. The subsequent equation can be applied to express the FoM [32].

$$FoM = \frac{WS}{FWHM} \ (RIU^{-1}) \quad (7)$$

In this context, *WS* signifies the wavelength sensitivity, and *FWHM* refers to the full width at half maximum, indicating the width of the loss spectrum at half of its peak value.

Sensor resolution delineates the sensor's capability to perceive deviations in the analyte's RI. This determination is conducted utilizing the following equation [33]:

$$R = \Delta n_a \times \frac{\Delta Wl_{min}}{\Delta Wl_{peak}} \ (RIU) \quad (8)$$

where, R is the sensor resolution, and ΔWl_{min} represents the smallest possible wavelength resolution.

3. Simulation Results and Discussion

3.1. Quantitative Findings and Sensitivity Evaluation

The evanescent electromagnetic field induced by light traveling within the fiber core is essential in a PCF-based SPR sensor, as it interacts with electrons on the metallic film and enables the excitation of a surface plasmon wave. The optical field is confined and propagates along the core region during light transmission, while the surface plasmon polariton (SPP) mode is generated

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at the sensing zone, where the plasmonic coating and overlayer are introduced. The operational characteristics depend on the integration of the core mode with the SPP mode. As the first step, we examine the evanescent electromagnetic field profile, which shows a significant coupling between the fundamental mode and the SPP mode for both transverse polarizations (x and y), as presented in Fig. 2 (a)-(d).

Fig. 2 (e-f) shows that, for both transverse polarizations (x and y), the plasmonic layer can excite the SPP mode within a certain wavelength range and resonantly interact with the fundamental mode at different frequencies. For x- and y-polarizations, the loss peaks appear at wavelengths of 910 nm and 920 nm, with corresponding confinement losses of 12.95 dB/cm and 89.21 dB/cm, respectively. Overall, y-polarization yields a sharper peak profile and a higher loss peak. This indicates that the plasmonic resonance state is more noticeable in y-polarization and that the SPP mode is excited more effectively.

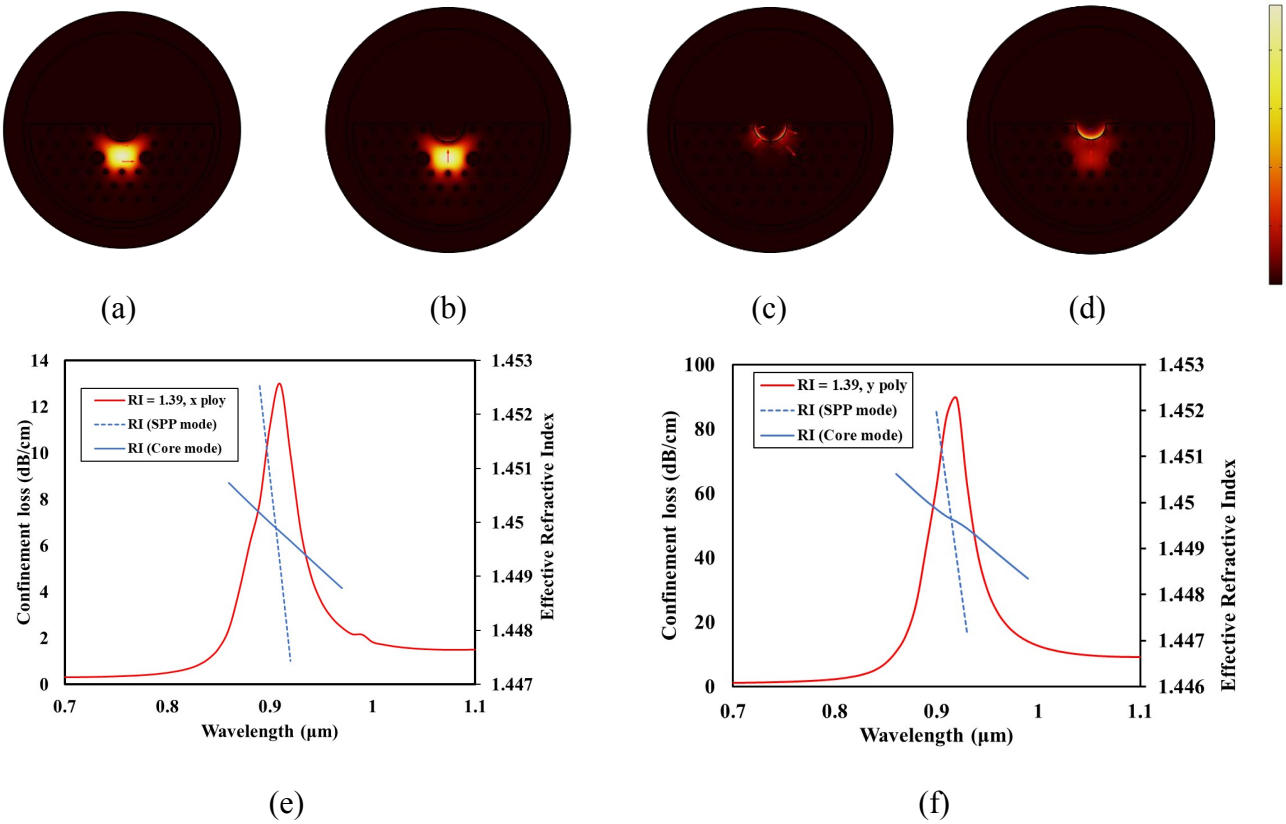

Fig. 2. The electric field profile of (a-b) the fundamental core mode and (b-c) the SPP mode, and dispersion curve of (e-f) confinement loss, fundamental mode, and SPP mode of n_a = 1.39, transverse polarizations (x and y), respectively.

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Since the majority of chemical sensing processes and biochemical interactions usually occur within the RI ranging from 1.33 to 1.42, this range is chosen for investigation [34], [35]. As the analyte RI fluctuates between 1.33 and 1.42, Fig. 3(a–b) displays the confinement loss characteristics for both transverse polarizations (x and y), which are used to evaluate the sensor's efficiency for multiple analytes. Increasing RI of the analyte induces the resonance peak to redshift toward higher wavelengths, as shown in Fig. 3 (a–b). A slight change in the RI of the analyte influences the plasmonic mode's RI, which in turn shifts the phase-matching position. When the analyte RI changes from 1.41 to 1.42, the resonance wavelength exhibits maximum shifts of 160 nm and 200 nm for transverse polarizations (x and y), respectively. The optimal wavelength sensitivities achieved at RI =1.41 of 16,000 nm/RIU and 20,000 nm/RIU for the transverse polarizations (x and y) modes, respectively.

The effectiveness of the sensor is further evaluated by determining amplitude sensitivity using the amplitude measurement approach for both transverse polarizations (x and y), as displayed in Fig. 3 (c–d). The y-polarization mode demonstrates greater performance (amplitude sensitivity) than the x-polarization mode, due to a more pronounced loss peak observed in the y-polarization response. The peak amplitude sensitivities of 653 RIU^{-1} and 1178 RIU^{-1} are measured for the transverse polarizations (x and y) modes, respectively.

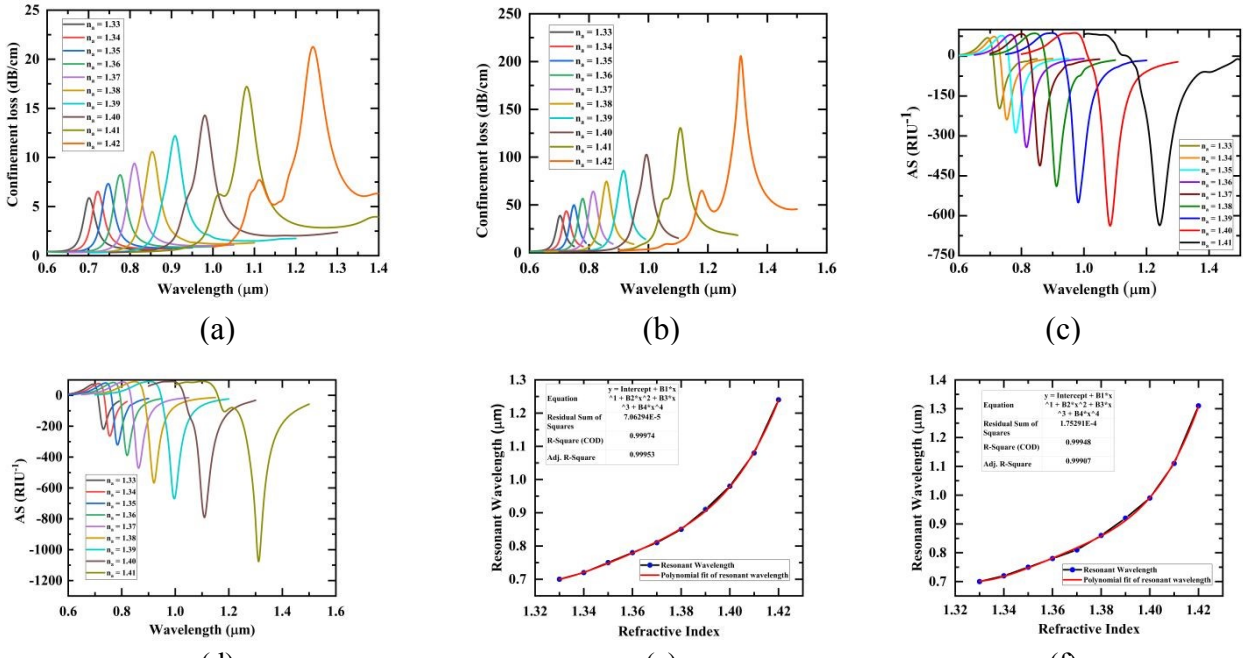

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Fig. 3. Dispersion profile of (a-b) confinement loss, (c-d) amplitude sensitivity, and (e-f) fitting polynomials to the resonance wavelengths for an analyte RI variation from 1.33 to 1.42 for transverse polarizations (x and y), respectively.

The sensor's performance is quantified using the figure of merit (FoM). A higher FoM signifies superior sensor performance. The estimated FoM values for various analyte RIs are presented in Table 3. The designed sensor obtains a maximum FoM of 265 and 366 for transverse polarization modes (x and y), respectively. Along with sensitivity and FoM, the resolution (R) of the designed sensor is evaluated to provide an in-depth assessment of its performance. The maximum resolutions of 6.25×10^{-6} and 5×10^{-6} (RIU) are observed for the transverse polarization modes (x and y), respectively.

The efficacy of a sensor can be assessed by examining the linearity of the fitting curves. A strong association between RI changes and sensor response is reflected by an increased slope of the curve fitting. Fig. 3 (e-f) illustrates the fluctuation in resonance wavelength and the corresponding fitting curves for analyte RI ranging from 1.33 to 1.42. The table displays the parameters and results for the fourth-order fitting.

The adjusted R^2 values for the fitted curve of the resonance wavelengths are 0.99953 and 0.99907 for transverse polarization modes (x and y), respectively, very close to 1. This indicates that the designed sensor demonstrates exceptional sensing capabilities.

Table 3. Assessment of sensor performance for amplitude sensitivity and wavelength sensitivity over different analyte refractive index variations.

Analyte RI	Resonant Wavelength (nm)		Wavelength Sensitivity (nm/RIU)		Amplitude Sensitivity (RIU^{-1})		FoM (RIU^{-1})	
	x-pol	y-pol	x-pol	y-pol	x-pol	y-pol	x-pol	y-pol
1.33	700	700	2000	2000	219	247	66	68
1.34	720	720	3000	3000	265	278	97	99
1.35	750	750	3000	3000	322	346	92	97
1.36	780	780	3000	3000	354	429	87	92
1.37	810	810	4000	5000	445	508	109	142
1.38	850	860	6000	6000	526	615	144	155
1.39	910	920	7000	7000	594	688	149	152
1.4	980	990	10000	12000	658	836	197	215
1.41	1080	1110	16000	20000	653	1178	265	366
1.42	1240	1310						

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In all cases, a more defined peak profile with a greater confinement loss is achieved in the y-polarization mode. This indicates that the SPP mode operates more effectively, and the plasmonic resonance state is more prominent in y-polarization. Therefore, in subsequent investigations, we examine the sensor's y-polarization reliability.

Table 4 presents a comparative study of the designed D-shaped PCF-based SPR sensor's performance with existing published PCF-based SPR sensors, emphasizing the key characteristics such as AS, WS, R, and FoM. Our designed D-shaped sensor exhibits excellent performance compared to the other sensors, offering the maximum AS of 1178 RIU^{-1}, the highest WS of 20000 nm/RIU, excellent wavelength resolution of 5×10^{-5} RIU, and a moderate FoM of 366 RIU^{-1} for y polarization mode. The aforementioned properties make the designed sensor exceptionally effective for accurate and precise refractive index detection.

Table 4. A comparative study of the proposed D-shaped PCF-based SPR sensor's performance with recently reported PCF-based SPR sensors.

Configuration	RI range	Max. WS (nm/RIU)	Max. AS (RIU^{-1})	R (RIU)	FOM (RIU^{-1})	Ref.
Dual-Channel D-Shaped	1.31–1.41	10,000	216	5×10^{-5}	125	[36]
D-shaped	1.35–1.46	5,600	_	8×10^{-4}	_	[37]
D-Shaped	1.26–1.38	5,400	_	_	_	[38]
Hoop-Cut	1.39–1.44	2,000	374.062			[39]
Proposed structure	1.33-1.42	20,000	1178	5×10^{-5}	366	

3.2. Variation of Architectural Parameters

This section highlights the sensor functionality by modifying structural parameters. The refractive indices (RIs) of 1.40 and 1.41 were selected as reference values to assess the sensor's performance under variations in geometric parameters. Subsequent investigations were carried out by adjusting these parameters while employing the selected RI values as benchmarks.

3.3. Influence of air space diameters on sensor performance

In a PCF, the air spaces act as the fundamental cladding zone. To enhance the sensor's sensitivity, the influence of geometric parameters was investigated by simulating various air space

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diameters (d_1, d_2). The initial assessment focused on the cladding air space (d_1), as depicted in Fig. 1 (a). Fig. 4 (a) demonstrates the loss peak of the structure for RIs of analyte ranging from n_a = 1.40 to n_a = 1.41. The data indicate that the loss peak increases as d_1 enlarges from 1.1 µm to 1.5 µm, as increasing the diameter of the smaller air space enhances the refractive contrast between cladding and core, improving phase matching between the SPP and core mode. The findings indicate that deviations of d_1 from its optimal diameter alter the loss characteristics, thereby affecting the AS, as illustrated in Fig. 4 (b). Table 4 shows that variations in d_1 do not affect the resonance wavelength, caused by the strong confinement of the y-polarized mode within the core region. As indicated by Fig. 4 (a-b) and Table 5, the sensor achieves its peak AS of 1178 nm/RIU at d_1=1.1 µm; therefore, this value was selected as the optimum diameter for the cladding small air space.

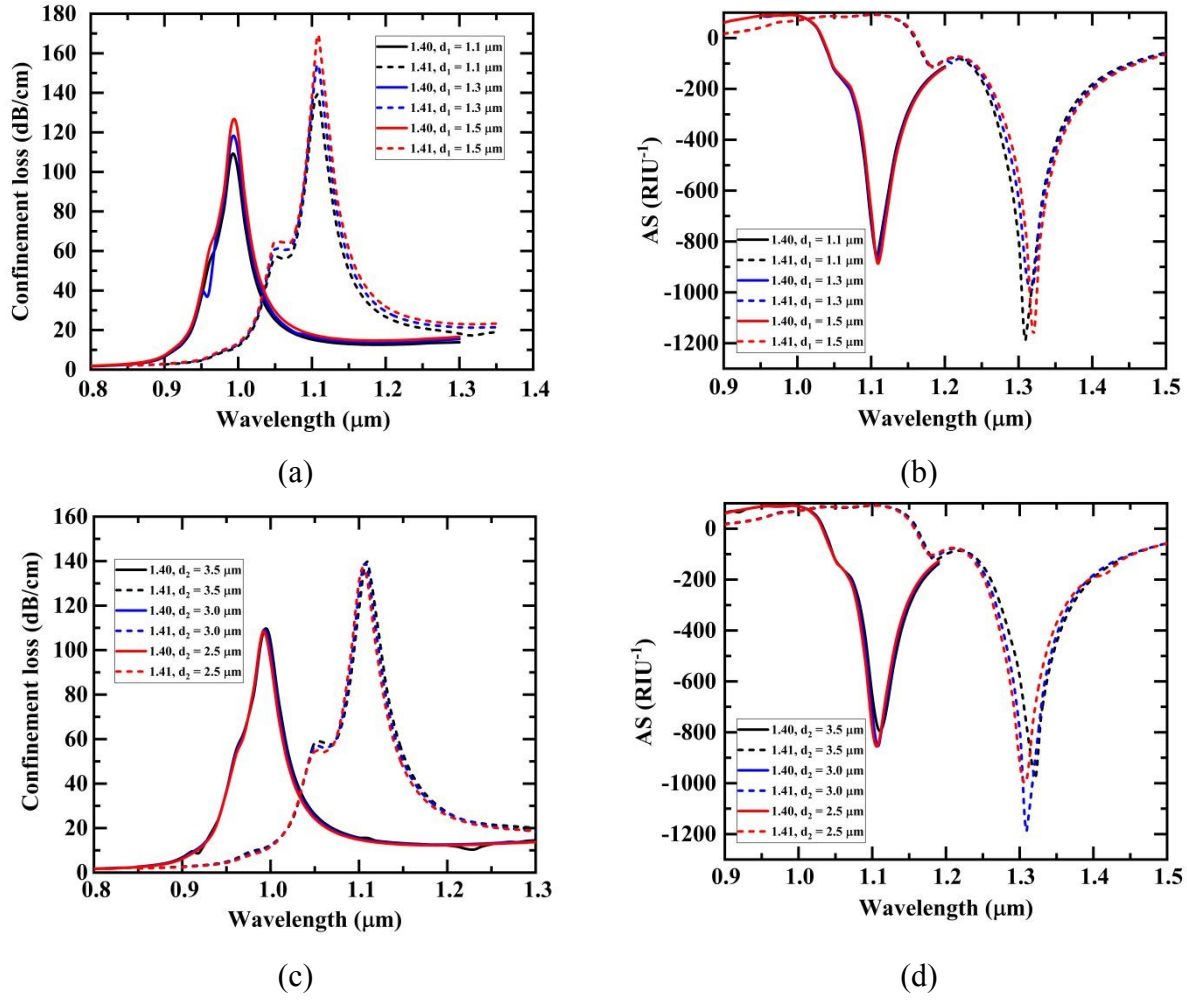

Fig. 4. Effect of the variation in air hole diameter d_1 and d_2 on the sensor's (a-c) confinement loss (b-d) AS for analyte RI varying from 1.40 to 1.41, respectively.

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Simulations were carried out for d_2 = 3.5, 3, 2.5 μm, with the outcomes displayed in Fig. 4 (c-d) and Table 5. Fig. 4 (c) shows that the peak loss for analyte RIs lies within the range of 1.40 to 1.41. The study indicates that the loss peak decreases as d_2 increases, and the loss peak exhibits an upward shift as d_2 decreases. The phenomenon occurs because of a bigger air space diameter, which effectively confines the core-directed mode, keeping more of the light within the core and reducing interaction with the metallic film. As a result, the excitation of surface electrons weakens, lowering the evanescent wave's penetration and the overall confinement loss. A red shift is also observed with the increase in space size. Fig. 4 (d) presents the amplitude sensitivity (AS) for n_a = 1.40 to n_a = 1.41. The overall best results were achieved at d_2 = 3 μm.

3.4. Influence of micro-channel on sensor efficacy

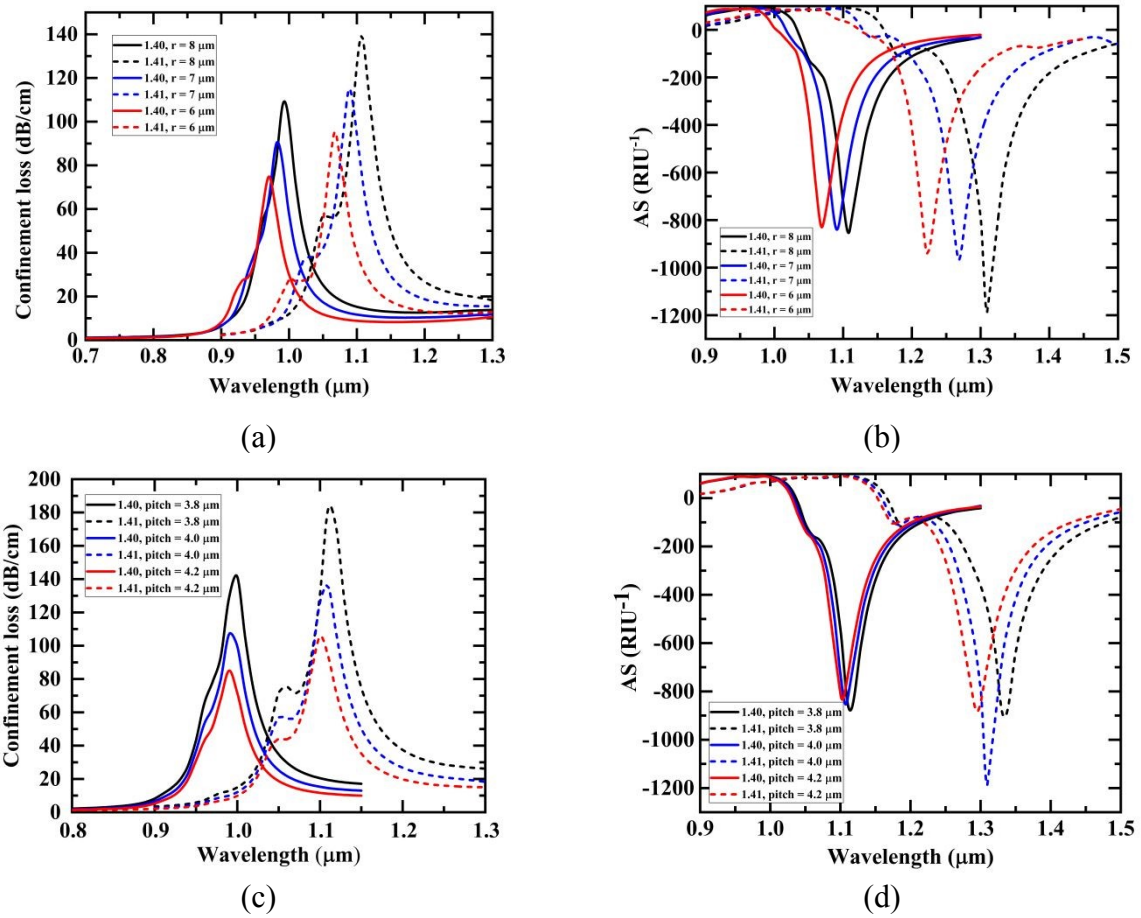

Fig. 5. Effect of the variation in microchannel diameter r and pitch distance on the sensor's (a-c) confinement loss, (b-d) AS for analyte RI varying from 1.40 to 1.41, respectively.

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The integration of a micro-channel in a D-shaped PCF-based SPR sensor improves analyte accessibility and boosts the interaction between guided and surface plasmon modes, resulting in increased resonance depth and sensitivity. Fig. 5 (a) explains the effect of varying the micro-channel diameter. The micro-channel diameter increases from 6 μm to 8 μm in increments of 0.5 μm. At r = 8 μm, the confined loss escalates markedly from 106.88 dB/cm to 135.83 dB/cm as the RI changes from 1.40 to 1.41, indicating a pronounced sensitivity of loss to small RI shifts in this range. Fig. 5 (b) illustrates that AS fluctuates considerably with changes in the value of r. The amplitude sensitivity values for 6, 7, and 8 μm are 913.7, 958.6, and 1178 RIU^{-1}at n_a = 1.41, respectively. Thus, 8 μm is selected as the optimum micro-channel diameter for the designed sensor. A detailed summary of the relationship between open channel diameter and sensor sensitivity is presented in Table 5.

Table 5. The optimized values for analyte detection range from 1.40 to 1.41.

Diameter (μm)	Wl_{peak} (μm) n_a=1.40	Wl_{peak} (μm) n_a=1.41	Confinement loss (dB/cm), n_a=1.40	Confinement loss (dB/cm), n_a=1.41	AS (RIU^{-1}), n_a=1.40	AS (RIU^{-1}), n_a=1.41
d_1						
1.1	0.99	1.11	106.88	135.83	836	1178
1.3	0.99	1.11	114.95	151.69	862	971
1.5	0.99	1.11	122.12	166.71	885	1164
d_2						
3.5	0.99	1.11	105.41	139.10	795	977
3	0.99	1.11	106.88	135.83	836	1178
2.5	0.99	1.1	107.06	130.99	824	977
r						
8	0.99	1.11	106.88	135.83	836	1178
7	0.98	1.09	88.59	114.80	838	958
6	0.97	1.07	74.79	93.42	826	913
Pitch (μm)						
3.8	1.00	1.11	141.45	182.23	842	875
4.0	0.99	1.11	106.88	135.83	836	1178
4.2	0.99	1.10	84.98	105.51	808	848

3.5. Impact of pitch on sensor effectiveness

Pitch of the air spaces influences the sensing efficacy of the sensor. In this study, the effect of the pitch variation is also investigated. To achieve optimal sensing performance, we precisely adjusted the pitch distance by gradually increasing it from 3.8 to 4.2 μm. Fig. 5 (c) illustrates that the peak confinement loss is achieved at pitch = 3.8 μm, about 141.45 and 182.23 dB/cm for the

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analyte RI ranging from 1.40 to 1.41, respectively. As the distance increases, the peak loss decreases because the effective mode index differences between the cladding and core increase. Fig. 5 (d) depicts that the amplitude sensitivity obtained for the $\Lambda = 4$ μm is higher than that of the $\Lambda = 3.8$ μm and 4.2 μm. The amplitude sensitivity of 875.53, 1178 , and 847.95 $RIU^{-1}$are achieved for the pitch distances $\Lambda = 3.8$, 4, and 4.2 μm, respectively, at n_a=1.41. Hence, 4 μm is picked as the most efficient pitch distance for the designed sensor. A detailed summary of the relationship between open channel diameter and sensor sensitivity is presented in Table 5.

3.6. Impact of gold (Au) coating depth on sensing efficacy

The gold layer height significantly influences the sensor's effectiveness, as it directly alters the resonance wavelength shift. This effect was assessed by examining variations in Au thickness. Fig. 6 (a) indicates that while the field confinement reduces with increasing Au thickness, the resonance peak experiences only a minor shift towards higher wavelengths for analyte RIs n_a = 1.40 to n_a = 1.41. To investigate the contribution of Au layer height on the confinement loss and the AS, the Au layer height was adjusted between 50 and 60 nm. Observations show, the loss is higher at Au = 50 nm, and lower at Au = 60 nm for analyte RIs n_a = 1.40 to n_a = 1.41, respectively. Fig. 6 (b) shows that the AS obtained for the 50 nm Au layer is higher than that of the 55 nm and 60 nm layers, whereas its WS is lower than that of the 55 nm layer. The 55 nm Au layer exhibits moderate confinement loss along with enhanced AS and WS. Thus, 55 nm is chosen as the optimum Au layer height for the designed sensor. A detailed summary of the correlation between Au layer height and AS is presented in Table 6.

3.7. Impact of TiO_2 coating depth on sensing efficacy

A thin TiO_2 dielectric overlayer modifies the plasmon–analyte interaction by enhancing the propagation of the evanescent electromagnetic field into the analyte medium, thereby improving sensor sensitivity. The confinement loss curve and AS responses for different TiO_2 layer heights are explained in Fig. 6 (c–d). To evaluate the contribution of TiO_2 height on the sensing characteristics, the layer thickness was gradually changed from 10 to 8 nm for analyte RIs extending from 1.40 to 1.41. A wavelength shift of 120 nm was seen in the loss spectrum peak with varying refractive index at a height of 9 nm for TiO_2, as presented in Fig. 6 (c), yielding the

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peak WS of 12000 nm/RIU. Fig. 6 (d) indicates that the amplitude sensitivity achieves its highest value of 1243 RIU^{-1} at a TiO_2 height of 10 nm, although its wavelength sensitivity is comparatively lower than that of the 9 nm layer. The 9 nm TiO_2 layer exhibits an optimal trade-off between confinement loss and sensitivity enhancement, exhibiting both superior wavelength and amplitude sensitivities with moderate confinement loss. Therefore, a TiO_2 layer height of 9 nm is identified as the efficient design parameter for the designed sensor. In Table 6, the correlation between the height of the TiO_2 layer and the associated sensing performance is summarized in detail.

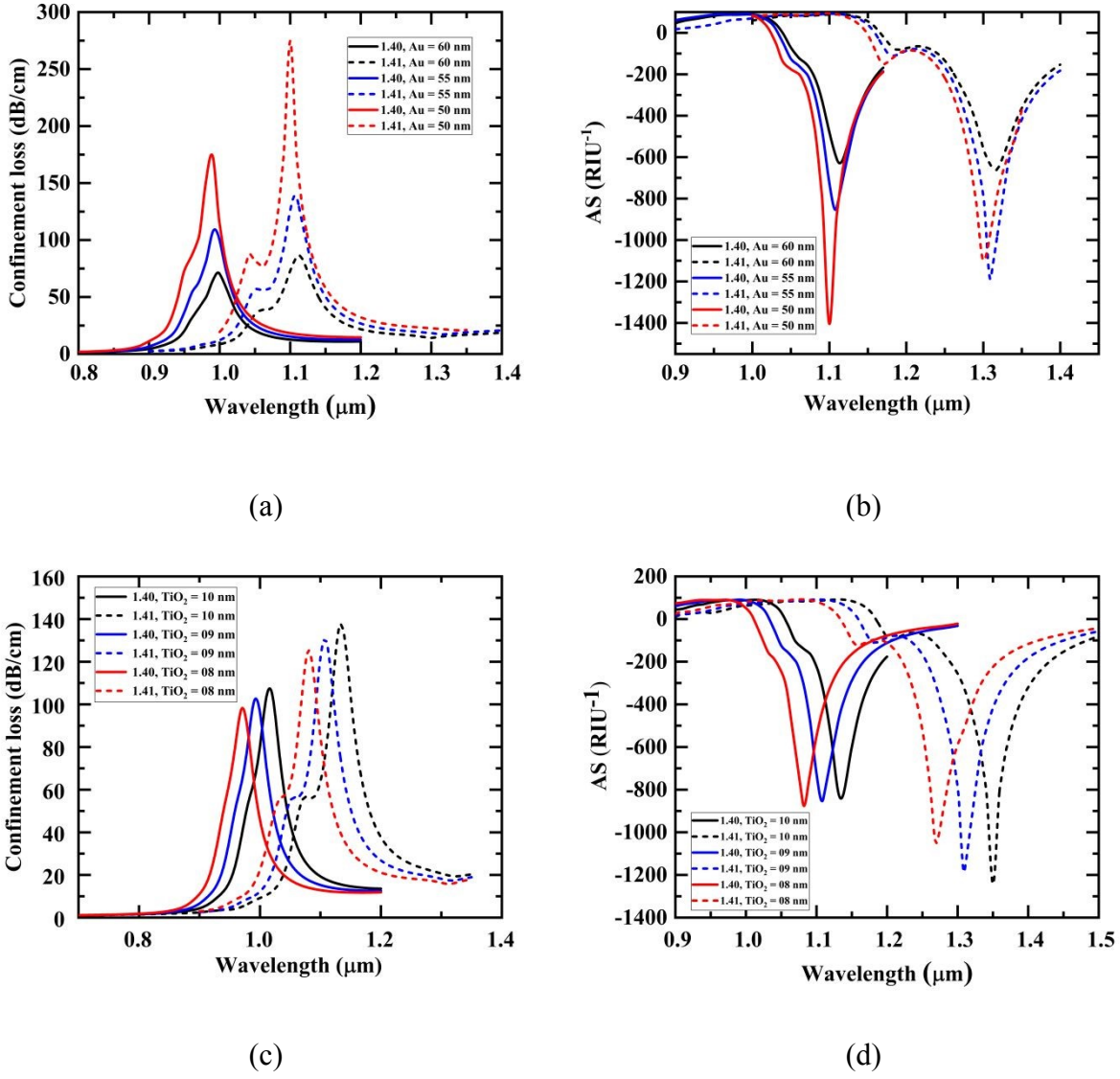

Fig. 6. Impact of the variation in Au and TiO_2 layer height on the sensor's (a-c) confinement loss, (b-d) AS for analyte RI varying from 1.40 to 1.41, respectively.

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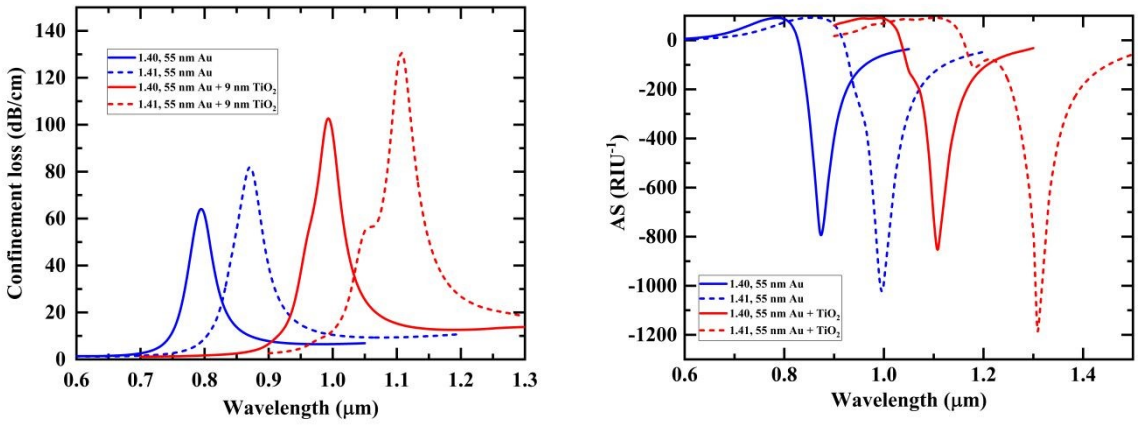
Table 6. A thorough explanation of the correlation between the Au and TiO_2 layer height with sensor sensitivity.

Material thickness (nm)	Wl_{peak} (µm) n_a=1.40	Wl_{peak} (µm) n_a=1.41	Confinement loss (dB/cm), n_a=1.40	Confinement loss (dB/cm), n_a=1.41	AS (RIU^{-1}), n_a=1.40	AS (RIU^{-1}), n_a=1.41
Au						
60	1.0	1.11	70.33	85.68	612.74	648.65
55	0.99	1.11	106.89	135.83	836.4	1178
50	0.99	1.10	173.75	274.77	1404.69	1099.65
TiO_2						
10	1.02	1.13	110.60	141.70	797.99	1243.24
9	0.99	1.11	106.87	135.83	836.4	1178
8	0.97	1.08	104.43	133.22	851.54	1050.49

3.8. Adding Only Gold Coating

The simulation uses an Au layer without a TiO_2 overlayer. The findings are displayed in Fig. 7 (a-b) and Table 7. Without an overlayer in the PCF-based SPR sensor design, the peak WS, AS, and FoM reached 12,000 nm/RIU, 987.10 RIU^{-1}, and 219.78 RIU^{-1} at n_a = 1.41, respectively. Although this shows moderate sensitivity, adding an overlayer further improved performance. The overlayer optimizes resonance by enhancing the connection between SPP and guided modes, increasing sensitivity. After adding a TiO_2 overlayer, the sensor achieved the ultimate WS, AS, and FoM of 20,000 nm/RIU, 1178.3 RIU^{-1}, and 366 RIU^{-1} at n_a = 1.41, respectively. This underscores the importance of overlayers in refining the sensor's optical characteristics for better detection efficiency and sensitivity.



(a) (b)

Fig. 7. Dispersion profile of (a) confinement loss and (b) AS for an analyte RI varied from 1.40 to 1.41, when the simulation uses only an Au layer and with a TiO_2 overlayer.

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Table 7. Evaluation of sensor performance for AS and WS over different analyte refractive index variations when the simulation uses only an Au layer and with a TiO_2 overlayer.

Analyte RI	WS (nm/RIU)		AS (RIU^{-1})		FoM (RIU^{-1})	
	Au	Au+TiO_2	Au	Au+TiO_2	Au	Au+TiO_2
1.33	2000	2000	99.80	247	49.75	68
1.34	1000	3000	132.39	278	26.04	99
1.35	2000	3000	172.25	346	54.35	97
1.36	2000	3000	239.64	429	55.56	92
1.37	3000	5000	323.84	508	84.03	142
1.38	4000	6000	430.69	616	109.59	155
1.39	5000	7000	592.80	689	127.88	152
1.4	8000	12000	758.35	836	178.57	215
1.41	12000	20000	987.10	1178.3	219.78	366
1.42						

3.9. Investigation of Fabrication Tolerance

These tolerance variations represent practical uncertainties arising during micro-channel formation and Au/TiO_2 thin-film deposition in real fabrication processes. Despite the advancement in the fabrication process, there is still a statistical chance of an inevitable uncertainty that could alter the design variables from their optimal value. So, it is essential to analyze the design tolerance level of the suggested sensor. Therefore, within this segment, the analyte RI value 1.39 is chosen to assess the fabrication tolerance with ± 2% variation of the sensor parameters. Fig. 8 (a) illustrates the confinement loss curve due to ± 2% variation of the pitch from its optimal value. With 2% decrease in the pitch from the optimal level, the loss increases, and with 2% increment, the loss decreases. However, shows no variation in the resonant wavelength of 920 nm in both cases. Fig. 8 (b) shows that ± 2% variation in the Au film thickness (t_{au}) from its optimum value influences the loss curve. As demonstrated in Fig. 8 (b), the resonant wavelength remained unchanged at 920 nm, and confinement loss varies as the thickness changes by ± 2% from its original value. The confinement loss grows from 89.21 to 95.17 dB/cm, by 2% decrease in gold thickness, and by 2% increase, loss decreases from 89.21 to 82.79 dB/cm. Furthermore, due to ± 2% deviation of the air space d_1 and d_2, there is no significant movement in the loss curve, and the loss curve is almost identical, depicted in Fig. 8 (c-d), respectively. From this study, the resonant wavelength remained unchanged at 920 nm for analyte RI 1.39, with only minor variations in confinement loss. This

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confirms that the proposed sensor is highly robust against ±2% variations in structural parameters and that the simulated performance can be reliably translated to practical fabrication scenarios.

Despite the promising numerical performance, especially the sensitivity of the proposed sensor, certain limitations should be acknowledged. This study does not consider the effects of other practical factors, such as temperature variations, measurement noise, and spectral broadening are not included in the simulations. These aspects can be evaluated in future work.

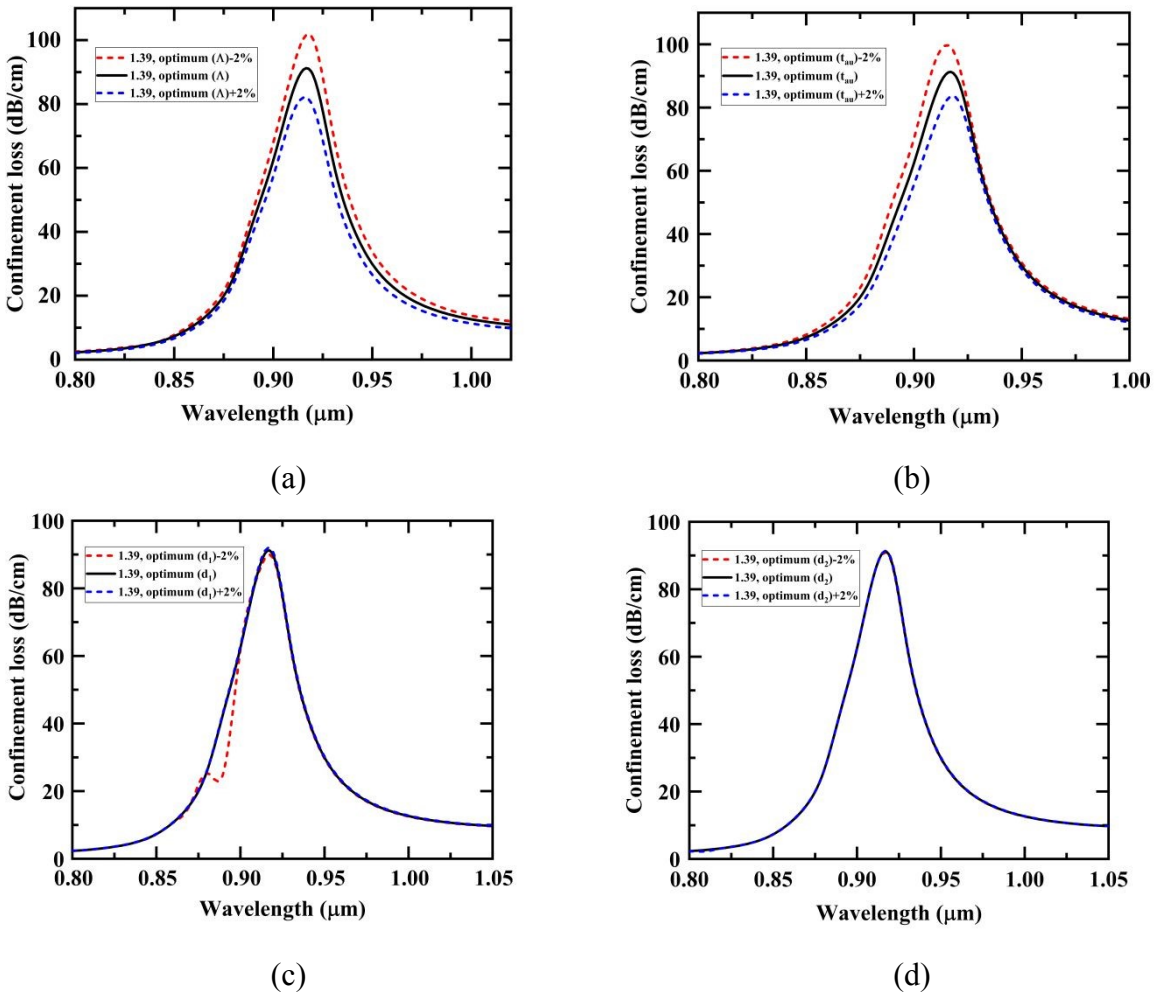

Fig. 8. Fabrication tolerance investigation for ± 2% deviation of (a) pitch (Λ), (b) Au layer thickness (t_{au}), (c) diameter of air space (d_1), and (d) diameter of air space (d_2).

3.10. Evaluation of the fabrication feasibility of the proposed PCF sensor

The fabrication viability of PCFs is investigated in this section, with a focus on the possibility of developing a particular fiber structure. Compared to conventional fibers, PCFs are

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more beneficial; several methods are utilized for the fabrication of the sensor, including stack-and-draw procedures, die-casting technique, and sol-gel processing [40]. The fabrication procedures for the micro-channel-based D-shaped sensor are demonstrated in Fig. 9 (a). Complex shapes such as circular, triangular, and square lattices can be easily created using the stack-and-draw procedure, which offers an extensible and versatile approach for the fabrication of PCFs. In this structure, thin and thick capillaries are employed to generate large and tiny air spaces, wherein guiding cores are created by swapping a capillary with a solid rod and then removing these rods entirely [41], [42]. After the fabrication of PCF, TiO_2, and Au layers on the PCF can be easily coated through various deposition techniques, such as the Chemical Vapor Deposition method and Atomic Layer Deposition (ALD) [43], [44]. The unidentified analyte can be detected by analyzing the confinement loss of the sensor.

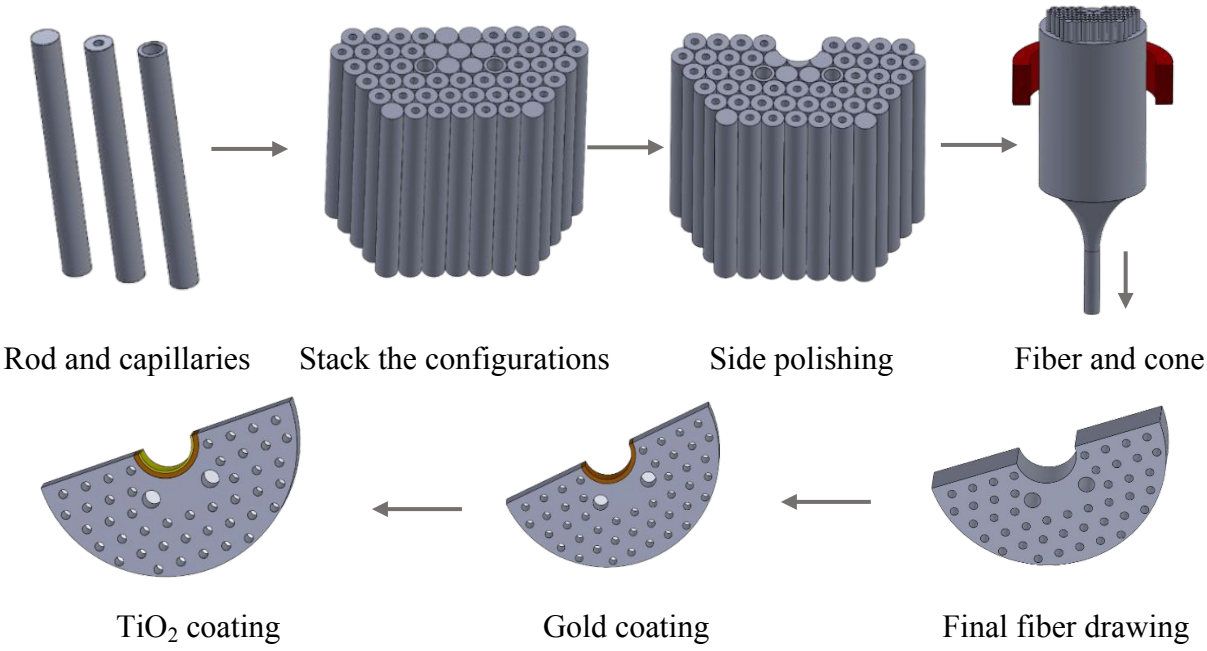

Fig. 9. Schematic of the designed sensor fabrication process.

Table 8. The RI variation for Healthy cells and Cancerous cells.

Cell Name	Cell Type and Concentration Level	RI		Ref.
		Healthy cell	Cancerous cell	
Basal cell	(30–70%) Non-malignant cell (80%) Malignant skin lesion	1.360	1.380	[45], [46]
HeLa	(30–70%) Non-malignant cell	1.368	1.392	[45], [46]

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	(80%) Cervical carcinoma			
Jurkat	(30–70%) Non-malignant cell (80%) Blood malignancy	1.376	1.390	[45], [46]
PC12	(30–70%) Non-malignant cell (80%) Adrenal carcinoma	1.381	1.395	[46], [47]
MDA-MB-231	(30–70%) Non-malignant cell (80%) Mammary carcinoma	1.385	1.399	[46], [47]
MCF-7	(30–70%) Non-malignant cell (80%) Mammary carcinoma	1.387	1.401	[46], [47]

3.11. Biomedical Application

This segment presents the optical responses at different wavelengths for both non-malignant and diseased biological samples, incorporating six different carcinoma cell types, utilizing the refractive indices from Table 8. The confinement loss performance of the D-shape PCF-based SPR sensor is shown in Fig. 10 (a–f) for a variety of biological samples, such as Basal, HeLa, Jurkat, PC12, and mammary carcinoma cell types MDA-MB-231 and MCF-7. The red line in every scenario denotes malignant cells, whereas the black line shows normal (healthy) cells. These differences in refractive index, originating from cellular variations such as cytoplasmic density and membrane composition, produce distinct plasmonic resonance shifts, enabling clear discrimination between healthy and malignant cells.

In every scenario, carcinoma cells show higher confinement losses and redshifted resonance peaks in comparison to their normal counterparts. One of the most frequently employed means for assessing sensor efficacy is spectral interrogation, which is based on the variations in resonance peaks across the wavelength spectrum. This idea can be clarified by applying equation (5), which examines wavelength changes associated with the loss peak and refractive index variations. Additionally, sensor effectiveness can be evaluated using the amplitude sensitivity at a particular wavelength, as calculated using Eq. (6). Based on the cancer cell types that were studied, the spectral sensitivities were 4000, 5000, 5714.29, 6428.57, 6428.57, and 7142.86 nm/RIU for Basal, HeLa, Jurkat, PC12, MDA-MB-231, and MCF-7 cells, while the corresponding amplitude sensitivities were 580.15, 628.57, 640.34, 669.29, 717.29, and 708.02 RIU^{-1}, respectively. Fig. 11 and Table 9 provide detailed information regarding the sensor's performance, including sensitivity, figure of merit (FOM), and resolution for six different types of carcinoma cells. The designed D-shaped sensor exhibits remarkable sensitivity, figure of merit, and resolution, signifying that this

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sensor delivers superior response. Moreover, it should be noted that all analyses, including cancer detection, are based on FEM simulations; experimental validation is the necessary next step to confirm the sensor's performance in real world sample detection.

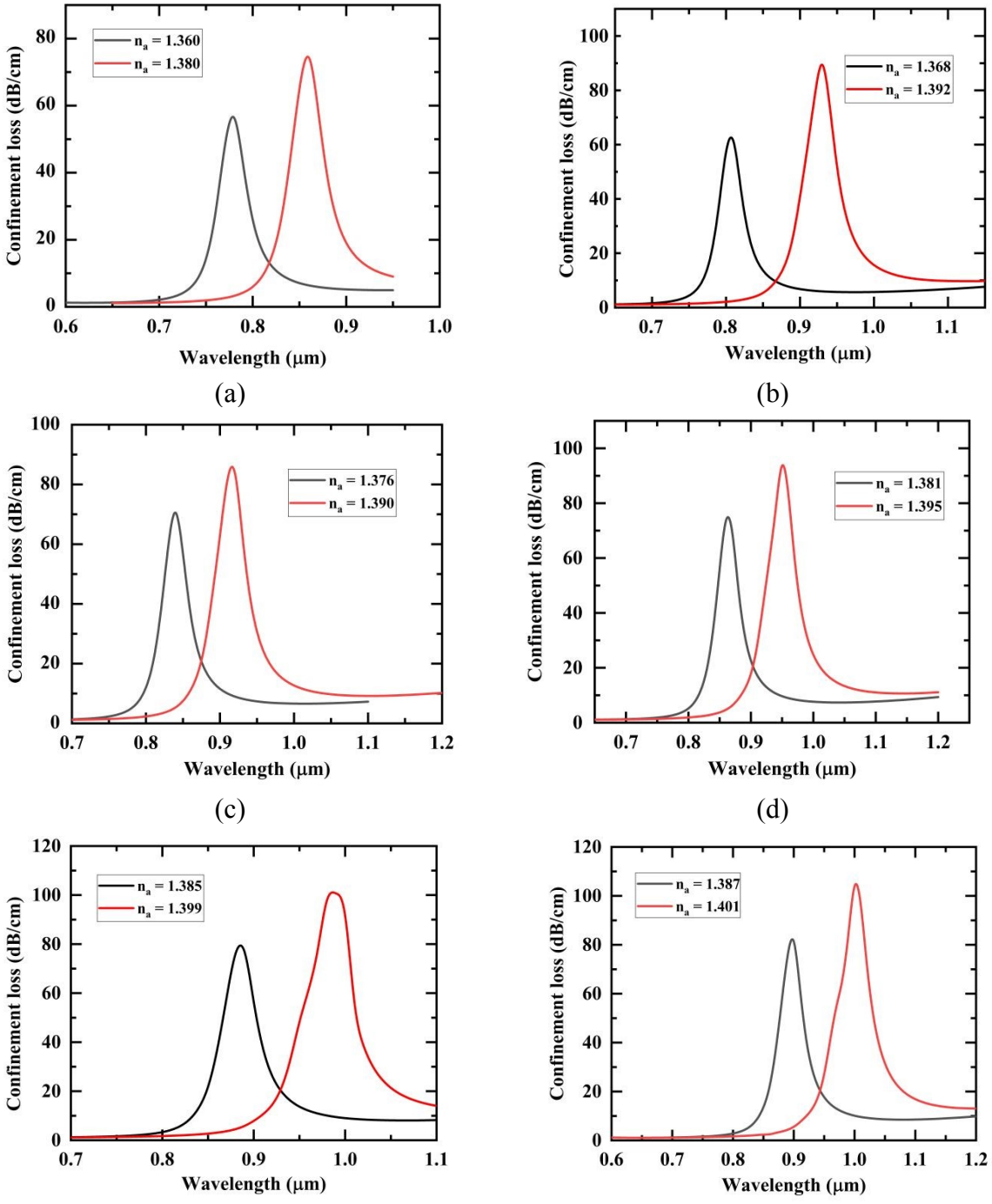

Fig. 10. (a-f) Outcomes of the identification of malignant skin lesions, cervical carcinoma, blood malignancy, adrenal carcinoma, and two variants of mammary carcinoma (MDA-MB-231 and MCF-7) utilizing the designed sensor.

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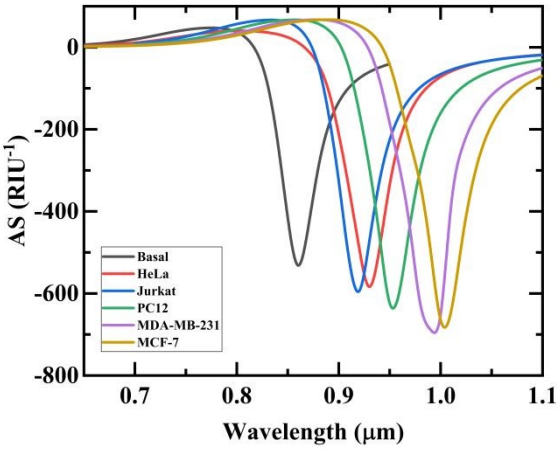

Fig. 11. AS for the following six carcinoma cell types: PC12, MDA-MB-231, Jurkat, HeLa, Basal, and MCF-7.

Table 9. Assessment of the Sensor's effectiveness at refined design variables for identifying different carcinoma cell types.

Cancer type	Cell type	WS (nm/RIU)	AS (RIU^{-1})	FOM (RIU^{-1})	Resolution (RIU)
Skin	Basal	4000.00	580.15	122.70	2.5×10^{-5}
Cervical	HeLa	5000.00	628.57	144.93	2.0×10^{-5}
Blood	Jurkat	5714.29	640.34	155.28	1.75×10^{-5}
Adrenal gland	PC12	6428.57	669.29	159.91	1.56×10^{-5}
Breast	MDA-MB-231	6428.57	717.29	153.06	1.56×10^{-5}
Breast	MCF-7	7142.86	708.02	165.34	1.40×10^{-5}

4. Conclusion

Accurate and early detection of cancer plays a vital role in enhancing patient outcomes, and non-invasive diagnostic techniques are making significant advances to address this issue. Among various non-invasive optical tools, D-shaped PCF-based SPR sensors are at the forefront due to their excellent sensitivity and fabrication simplicity. This study has designed a PCF-based SPR sensor for the identification of various carcinoma cells utilizing a D-shaped structure. The spectral and amplitude interrogation methods have been employed to investigate the performance of different sensor configurations. The impact of incorporating multiple air holes and their size, diameter, pitch distance, along with the impact of adding a plasmonic layer, TiO_2 overlayer, and their thickness on sensor performance has been comprehensively analyzed to obtain the optimized

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structure. The sensor achieved the highest amplitude and spectral sensitivities of 717.29 RIU^{-1} and 7142.86 nm/RIU, respectively, recorded for MCF-7 and MDA-MB-231 cells. Additionally, the designed sensor exhibited the highest spectral sensitivity of 20,000 nm/RIU across the RI span of 1.330 to 1.420 for y-polarized light. The sensor also depicted outstanding FOM, AS, and resolution values of 366 RIU^{-1}, 1178 RIU^{-1}, and 5×10^{-6} RIU, respectively, within the aforementioned RI range. Finally, a convenient and effective way of the presented sensor's fabrication is proposed with a detailed analysis of fabrication feasibility via investigating its fabrication tolerance. The outcomes of the simulation show an impressive improvement in sensitivity and accuracy compared to the state-of-the-art studies in the literature, which is a testament that the designed architecture could be an effective tool for the precise diagnosis of cancer with better resolution. It should be noted that all analyses, including cancer-cell detection and fabrication-tolerance evaluation, are based on FEM simulations; experimental validation is required to confirm the sensor's performance in real biological environments.

CRediT authorship contribution statement

Kazi Zannatul Ferdwushee : Writing – original draft, Software, Formal analysis, Methodology, Visualization. Md Abu Shahid Chowdhury: Conceptualization, Supervision, Writing – original draft, Formal analysis, Validation, Visualization. Siddika Tamanna Islam: Writing – review & editing.

Conflicts of Interest

The authors declare they have no conflicts of interests.

Data availability statement

The data that support the findings of this study will be made available from the corresponding author upon reasonable request.

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Data Availability Statement

The data supporting this study are available from the corresponding author upon reasonable request.