Double-sided Polishing Photonic Crystal Fiber Biosensor Based on Surface Plasmon Resonance

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Double-sided polishing photonic crystal fiber biosensor based on surface plasmon resonance
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ABSTRACT

In this paper, we propose a high-sensitivity double-sided polishing photonic crystal fiber biosensor based on surface plasmon resonance. The performance of the sensor was analyzed by finite element method, including the influence of different pore sizes and position, polishing depth, pore spacing, metal thickness, refractive index(RI), and manufacturing error (5%) on the sensing performance. We also analyze the performance of the modified sensor in early cancer detection applications. Our use of a double-sided polishing structure and coating with a gold layer not only increases the contact area between the analyte and gold layer but also provides a large enough space for the analysis channel, which is conducive to the actual manufacture and use. The sensor has a detection range of 1.35-1.39, the maximum wavelength sensitivity of 21500 nm/RIU, the average wavelength sensitivity is 9550 nm/RIU, the amplitude sensitivity of -1057.1756 RIU⁻¹, the maximum resolution is 4.68×10^{-6} , R^2 of 0.98265, figure of merit is 250, and the maximum resolution of 4.68×10^{-6} , which can be used in the cancer detection and other biological areas.

Keywords: Fiber optic biosensor, Double-sided polishing photonic crystal fiber, Surface plasmon resonance, Cancer detection sensor.

1. Introduction

At present, surface plasmon resonance (SPR) sensors are used in a variety of fields, such as environmental gas detection, biochemical medical analysis, and food safety [1]. The sensor has the characteristics of real-time, high sensitivity, and remote control [2,3]. Kretschmann experimented with

SPRS in 1968, but it was expensive, bulky, and unreliable [4,5]. These shortcomings make PRISM SPR unable to distributed sensing and mass production [6,7].

To solve this problem, a photonic crystal fiber sensor based on surface plasmon resonance is developed by combining photonic crystal fiber with surface plasmon resonance (PCF-SPR) [8-10]. This kind of sensor has the advantages of flexible structure, anti-electromagnetic interference, high integration, and high mechanical stability [11,12]. In 2015, Sharma et al. proposed a photonic crystal fiber sensor for cancer detection [13]. In 2018, Liu et al. designed a D-shaped photonic crystal fiber biosensor with a wavelength sensitivity of 3340 nm/RIU [14]. In 2020, Yasli developed a photonic crystal fiber biosensor to detect six types of cancer single cells [15]. In the field of biosensors, since the refractive index of blood cells, cervical cells, skin basal cells, adrenaline, and breast cells are all in the range of 1.35-1.39 [16], it is meaningful to propose a biosensor with a detection range of 1.35-1.39. However, to obtain higher sensitivity, people have studied the D-shape PCF-SPR to carry out deeper polishing to increase the contact area with metal, which will lead to a sharp increase in transmission loss, which is not conducive to detection [17]. Some other PCF-SPR sensors are also difficult to be used in practical applications due to their impractical analyte channels.

This paper presents a PCF-SPR sensor with double-sided film polishing. It has both realizable analyte channels and low confinement loss. The maximum wavelength sensitivity of the sensor proposed in this paper is 21500 nm/RIU, the average wavelength sensitivity is 9550 nm/RIU, the amplitude sensitivity is -1057.1756 RIU⁻¹, and the maximum resolution is 4.68×10⁻⁶, which can realize the monitoring between RI 1.35-1.39, and can be applied in the biomedical field in practice.

2. The structure of the PCF and basic theory

Fig. 1 2D cross-sectional view of the PCF-SPR sensor.

Fig. 1 is a 2D cross-section of the PCF-SPR sensor model proposed in this paper. The diameter of air holes d_1 , d_2 and d_3 are 2.0 μm , 0.4 μm , and 1.0 μm respectively. We set Λ to 2.0 μm , spacing d_1 to 1.2 Λ , spacing d_2 to 1.6 Λ transverse, spacing 2.4 Λ longitudinal, and spacing d_3 to 2.8 Λ . The large air hole is arranged in a simple quadrilateral shape, which helps to reduce the production difficulty. The air hole d_2 is arranged in a rhombic shape, which can increase the coupling strength of y polarization. The air hole d_2 effectively limits the mode field position of the core mode. The h is the height from the PCF center to the casting mold position, and the gold layer thickness t is set to 40 nm. The gold layer can be coated by magnetron sputtering [18] or chemical vapor deposition [19]. The polishing mold can be polished with the edge arc glass base method and the polishing accuracy is 0.01 μm [20].

The dielectric constant of silica is calculated by Sellmeier equation [21]:

$$n^2(\lambda) = 1 + \frac{A_1\lambda^2}{\lambda^2 - B_1} + \frac{A_2\lambda^2}{\lambda^2 - B_2} + \frac{A_3\lambda^2}{\lambda^2 - B_3} \quad (1)$$

Where A_1 , A_2 , A_3 , B_1 , B_2 , and B_3 are 0.696166300, 0.407942600, 0.897479400, 4.67914826×10⁻³ μm^2 , 1.35120631×10⁻² μm^2 , and 97.9340025 μm^2 , respectively. There, λ is the incident light wavelength in a vacuum.

The wavelength dependence on the dielectric constant of the gold layer is calculated using the Drude-Lorentz model [22]:

$$\epsilon_{Au} = \epsilon_{\infty} - \frac{\omega_D^2}{\omega(\omega + j\gamma_D)} - \frac{\Delta\epsilon\Omega_L^2}{(\omega^2 - \Omega_L^2) + j\Gamma_L\omega} \quad (2)$$

Where γ_D and ω_D are the damping and plasma frequency, respectively. Here $\Delta\epsilon$ is the permittivity at a high frequency, having a value of 5.9673, and ϵ_{Au} represents the permittivity of Au. Additionally, other parameters in the model have the numerical value of $\omega_D/2\pi=2113.6$ THz, $\gamma_D/2\pi=15.92$ THz, and $\Delta\epsilon=1.09$. Note that, the spectral width and the oscillator strength is $\Gamma_L/2\pi=104.86$ THz and $\Omega_L/2\pi=650.07$ THz.

The confinement loss is used to analyze the characteristics of the sensor and can be evaluated by the following equation [23]:

$$Loss(dB/cm) \approx 8.86 \times \frac{2\pi}{\lambda} \times Im(n_{eff}) \times 10^4 \quad (3)$$

$Im(n_{eff})$ is the virtual part of the effective index of the fundamental mode, which can be obtained by COMSOL software, λ is the working wavelength.

In this paper, the sensitivity of the sensor is calculated by wavelength and amplitude interrogation methods. The formula of wavelength sensitivity is [24]:

$$S_{\omega}(nm/RIU) = \frac{\Delta\lambda_{peak}}{\Delta n_a} \quad (4)$$

In the formula, $\Delta\lambda_{peak}$ is the peak wavelength shift, and the n_a is the change of the refractive index (RI) of the analyte.

The amplitude sensitivity formula is [25]:

$$S_A(RIU^{-1}) = -\frac{1}{\alpha(\lambda, n_a)} \frac{\delta\alpha(\lambda, n_a)}{\delta n_a} \quad (5)$$

In the formula, $\alpha(\lambda, n_a)$ is the confinement loss at the n_a analyte RI, and $\delta\alpha(\lambda, n_a)$ is the difference of the confinement loss generated by two adjacent RI of the analyte.

The resolution equation is [26]:

$$R(RIU) = \frac{\Delta n_a \times \Delta\lambda_{min}}{\Delta\lambda_{peak}} \quad (6)$$

Among them, $\Delta\lambda_{min}$ is the resolution of the spectrometer, which is usually set at 0.1 nm.

Figure of merit(FOM) is an important parameter for measuring sensor performance by combining sensitivity and full-width-half-maximum(FWHM), which is defined as [27] :

$$FOM = \frac{S_{\omega}}{FWHM} \quad (6)$$

The PCF is usually connected to a single-mode fiber (SMF). At this point, a broadband light source is used to transmit the light into the PCF via the SMF. The sensing process is based on the coupling between the surface plasmon polariton(SPP) mode and the core mode. When the refractive index of the two effective modes is the same, a sharp loss peak will result in the coupling. When the refractive index of the analyte changes, the resonance wavelength changes accordingly. The PCF can sense this change and output it to a high-resolution spectrograph for analysis [28].

3. Simulation results and analysis

Fig. 2. (a) The dispersion relationship of core-guided mode, SPP mode and confinement loss spectrum at $RI=1.38$. (b) y-polarized core mode at $\lambda = 774$ nm (phase matching point); (c) y-polarized SPP mode at $\lambda = 760$ nm; (d) x-polarized core mode at $\lambda = 760$ nm; (e) y-polarized core mode at $\lambda = 750$ nm.

The coupling characteristics of the two-side polishing mold PCF-SPR sensor are the same as those of the common PCF-SPR sensor. Fig. 2 (a) shows the confinement loss spectra of the core modes in the x-polarization (x-pol), y-polarization (y-pol) and the dispersion relationship between the core model in the y-direction and the SPP mode when the refractive index of the analyte is 1.38. When the real part of the effective refractive index of the core mode is equal to that of the SPP mode, the two modes will be coupled to achieve the maximum energy transmission between the two modes, and the loss value will be the highest. The electric field diagram is shown in Fig. 2 (b). Fig. 2 (c) to (e) are the electric field diagrams of SPP mode, y-pol core mode, and x-pol core mode at the wavelength of 760 nm respectively. It can be seen that the loss value of the core mode of x-pol is extremely low. Therefore, subsequent studies are based on the core mode of y-pol.

Fig. 3. Sensor performance. (a) Confinement loss spectra from 1.35 to 1.39. (b) Amplitude sensitivity spectra from 1.36 to 1.39. (c) linear curve fitting of the resonant wavelength; (d) Wavelength sensitivity and FOM from 1.35 to 1.38.

The double-polishing PCF-SPR sensor presented in this paper has good sensing performance. As can be seen from Fig. 3 (a), with the increase of the refractive index of the solution to be measured, the

resonance wavelength is red-shifted and the limiting loss increases. The reason for this phenomenon is that when the RI refractive index of the liquid to be tested increases, the evanescent field becomes stronger, leading to stronger coupling. At the same time, the effective refractive index of SPP mode increases while the core model remains unchanged. Fig.3 (b) shows the amplitude sensitivity of the sensor, which is the largest when the liquid to be tested RI is 1.39 and is $-1057.1756 \text{ RIU}^{-1}$. Fig. 3 (c) is the linear polynomial fitting of resonance wavelength, $R^2=0.98265$, which proves that the sensor has a good continuous working capability. Fig. 3 (d) shows wavelength forest sensitivity and FOM, with the maximum wavelength sensitivity of 21500 nm/RIU and the maximum FOM of 250.

Table 1 Performance comparison of the sensor with recently reported sensors.

Fiber Structure	Refractive Index (RIU)	Amplitude Sensitivity (RIU^{-1})	Wavelength Sensitivity (nm/RIU)	Wavelength Resolution (RIU)	FOM	Published Year and Ref
Double core PCF	1.45-1.48	-700	8000	1.25×10^{-5}	138	2019[29]
Double sided polishing PCF	1.30-1.342	N/A	8000	1.3×10^{-5}	N/A	2019[30]
The pores are filled with metal PCF	1.454-1.478	N/A	8776.8	N/A	168.16	2020[31]
annular microfluidic channel PCF	1.33-1.36	-633.40001	10600	9.43×10^{-5}	303	2020[32]
Side-polished PCF	1.35-1.39	N/A	12500	8.0×10^{-6}	N/A	2021[33]
Curved polished PCF	1.36-1.402	-757	7142.86	3.2×10^{-5}	N/A	2021[15]
Our work	1.35-1.39	-1057.1756	21500	4.65×10^{-6}	250	

Table 1 shows the performance comparison between the proposed sensor and the sensor published in the last three years. It can be seen from Table 1 that the dual-polished PCF-SPR biosensor proposed in this paper has a better performance compared with other PCF-SPR biosensors, and the same performance is also due to the previously published dual-polished PCF-SPR biosensor.

Fig. 4. (a)-(c) Confinement loss spectra with varying silver thicknesses and (d)-(f) Amplitude sensitivity spectra with varying silver thicknesses.

The influence of metal film thickness on sensing performance is great. Fig.4 (a), (b), and (c)

respectively show the confinement loss spectra of RI from 1.35 to 1.39 when the thickness of the gold layer is 30 nm, 40 nm, and 50 nm. Fig. 4 (d), (e), and (f) is the amplitude sensitivity of the gold layer with 30 nm, 40 nm, and 50 nm thickness. It can be seen that with the increase of metal thickness, the confinement loss of the same refractive index decreases, and the amplitude of the amplitude sensitivity decreases. This is because the transfer of light from the core to the analyte is reduced by the thicker gold layer. However, the actual manufacturing and use will cause losses to the gold layer, so we choose the gold layer thickness of 40 nm.

Fig. 5. Confinement loss spectra with varying polishing depth h . (a) $h=2.35 \mu\text{m}$; (b) $h=2.40 \mu\text{m}$; (c) $h=2.45 \mu\text{m}$. (d) Polynomial linear fitting.

The depth of the polishing mode has a great influence on the dual-polishing photonic crystal fiber. As can be seen from Fig. 5 (a) to (c), with the decrease of polishing depth (h increased), the confinement loss of the core mode increases, and the resonance wavelength redshift, but the wavelength sensitivity decreases. This is because the distance between the core mode and the SPP mode decreases with the decrease of the polishing depth, and the coupling phenomenon is more likely to occur. Fig. 5 (d) is a polynomial fitting curve, R^2 decreases with the decrease of polishing depth. Combined sensitivity, confinement loss, and linear relationship, h was chosen as $2.40 \mu\text{m}$. This also proves that double-sided polishing PCF can solve the problem of high loss caused by the increase of polishing depth of D-shape PCF.

Fig. 6. Confinement loss spectra with varying d_1 . (a) $d_1=1.8$ um; (b) $d_1=2.0$ um; (c) $d_1=2.2$ um; (d) Polynomial linear fitting.

Fig. 7. Confinement loss spectra with varying d_2 . (a) $d_2=0.2$ um; (b) $d_2=0.4$ um; (c) $d_2=0.6$ um; (d) Polynomial linear fitting.

Then we analyze the influence of air holes on the sensing performance. Fig. 6 to Fig. 8 respectively analyze the influence of d_1 , d_2 , and d_3 of different sizes on the sensor performance. It can be seen from Fig. 6 and Fig. 7 that d_1 and d_2 have similar effects on the sensor. With the increase of diameter, the resonance wavelength is redshifted, the confinement loss is reduced and the sensitivity is increased. As the diameter of the two air holes increases, the coupling channel width between the core mode and the SPP mode decreases, making the coupling phenomenon more difficult to occur. Since the air hole d_2 is in the middle of the coupling channel, it has a greater impact on performance. Fig. 6 (d) and Fig. 7 (d) are polynomial linear fitting curves of different d_1 and d_2 respectively. The change of R^2 after the same air hole size can also prove that the d_2 size on the coupling channel has a greater impact on the sensing performance. Combined sensitivity and confinement loss, d_1 and d_2 are set to 2.0 um and 0.4 um, respectively.

Fig. 8. Confinement loss spectra with varying d_3 . (a) $d_3=0.8 \mu\text{m}$; (b) $d_3=1.0 \mu\text{m}$; (c) $d_3=1.2 \mu\text{m}$; (d)Polynomial linear fitting.

From Fig. 8 (a) to (c), it can be seen that the diameter of air hole d_3 does not influence sensor performance. That's because it has no contact-coupled channel. However, air hole d_3 can limit the location of the core mode and increase the fault tolerance of the fabrication.

Fig. 9. (a) Confinement loss spectra with varying pitch size; (b) Confinement loss spectra with varying size error.

Fig. 9 (a) shows the confinement loss spectra with varying pitch sizes. It can be seen that with the increase of the interval, the resonance wavelength shifts blue, the loss increases, and the sensitivity changes little. This is due to the increase of spacing, the coupling channel between the core mode and the SPP mode becomes wider, and the coupling phenomenon is more likely to occur. Generally, the manufacturing error of photonic crystal fiber is within 5%. Fig. 9 (b) and (c) show the loss spectrum with plus or minus 5% error (Fiber air hole size) at RI of 1.37. It can be seen that when the size error exists in the fabrication of photonic crystal fiber, the influence on the resonance wavelength is small, but the influence on the confinement loss is large.

Fig.10. (a) Confinement loss spectra with a skin cell and blood cell; (b)Amplitude sensitivity.

The biosensor proposed in this paper can be applied in the field of cancer detection. Fig. 10 shows the sensing performance of this sensor for specific applications in the cancer detection field. According to the existing literature, the refractive index of normal skin cells is 1.36, after canceration, the refractive index of cells was 1.38. The refractive index of normal blood cells is 1.376, and after canceration, the refractive index of cells was 1.39 [16]. The sensor can be applied to detect these two cancerous phenomena with a sensitivity of 6500 nm/RIU and 17875 nm/RIU and amplitude sensitivity of -642.60477 RIU⁻¹ and -1478.69277 RIU⁻¹ respectively. It has very superior performance.

4. Conclusions

In conclusion, a biosensor with double-sided polishing film PCF-SPR coated with the nanoscale gold film is proposed, and the dispersion relationship and modal loss analysis are carried out by using the finite element method. The performance of the biosensor was analyzed by using wavelength and amplitude interrogation methods. The maximum wavelength sensitivity is 21500 nm/RIU, the average wavelength sensitivity is 9550 nm/RIU, the amplitude sensitivity is $-1057.1756 \text{ RIU}^{-1}$, R^2 is 0.98265, FOM is 250 and the maximum resolution is 4.68×10^{-8} , which can realize the monitoring between RI 1.35–1.39. It has a wavelength sensitivity of 17875 nm/RIU and the amplitude sensitivity of $-1478.69277 \text{ RIU}^{-1}$ in the detection of the blood cancer cell. The sensor has excellent performance and can be applied to single cell detection of cancer.

Declarations

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Conflicts of interest. The authors declare no conflicts of interest.

Availability of data and material. The datasets generated during and/or analysed during the current study are available from the corresponding author on reasonable request.

Code availability. Not applicable.

Ethics approval. Not applicable.

Authors' contributions. All authors contributed to the study conception and design. Material preparation, data collection and analysis were performed by Zhiwen Zhang. The first draft of the manuscript was written by Zhiwen Zhang and all authors commented on previous versions of the manuscript. All authors read and approved the final manuscript.

Consent to participate. Not applicable.

Consent for publication. Not applicable.

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