

Intensity-modulated TFBG–SPR biosensor for rapid label-free detection of human IgG

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Received May 6, 2026 | Accepted June 17, 2026 | Published Online August 6, 2026

Tilted fiber Bragg grating (TFBG)-based surface plasmon resonance (SPR) sensors have attracted considerable attention for optical fiber biochemical sensing owing to their compact structure and strong cladding-mode interaction. However, the limited biomolecule immobilization efficiency on metal surfaces often restricts sensing performance. In this work, a TFBG–SPR biosensor functionalized with a polydopamine–graphene oxide (PDA–GO) composite layer is proposed for rapid and label-free detection of human immunoglobulin G (IgG). A femtosecond laser-inscribed TFBG coated with a thin gold film is employed to excite SPR, while the PDA–GO layer provides a stable biofunctional interface for biomolecule interaction on the sensing surface. An intensity-based interrogation scheme is implemented by monitoring the intensity variation of a selected cladding mode within the SPR attenuation band. Experimental results demonstrate a clear concentration-dependent response to IgG within the tested range of 20–60 $\mu\text{g}\cdot\text{mL}^{-1}$, with an exponential fitting coefficient of $R^2 = 0.997$ and an average intensity sensitivity of 0.0475 dB $(\mu\text{g}\cdot\text{mL}^{-1})^{-1}$. The sensing signal reaches equilibrium within approximately 1 min, indicating rapid detection capability. The proposed PDA–GO-functionalized TFBG–SPR sensor provides a compact and effective platform for label-free biochemical detection and shows promising potential for optical fiber biosensing applications.

Keywords: graphene oxide; human immunoglobulin G; optical fiber biosensor; polydopamine; surface plasmon resonance; tilted fiber Bragg grating.

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DOI: [10.3788/LPB.2026.20007](https://doi.org/10.3788/LPB.2026.20007)

1. Introduction

Surface plasmon resonance (SPR)-based optical fiber biosensors have attracted considerable attention in recent years because they combine label-free detection, compact structure, remote interrogation, and compatibility with miniaturized sensing platforms^[1–4]. Compared with traditional prism-based SPR systems, optical fiber SPR sensors offer important advantages, including small size, flexible deployment, and suitability for in situ biochemical detection^[5–7]. Among various fiber-based sensing configurations, tilted fiber Bragg gratings (TFBGs) have emerged as a powerful platform for plasmonic sensing due

to their unique ability to couple the core mode to a series of backward-propagating cladding modes^[8–10]. When a thin metal layer is deposited on the fiber surface, specific cladding modes can satisfy the phase-matching condition with surface plasmon waves at the metal–dielectric interface, producing a pronounced attenuation band in the transmission spectrum^[11].

Owing to the strong evanescent-field interaction between cladding modes and the surrounding medium, TFBG–SPR sensors exhibit high refractive-index sensitivity and excellent potential for biochemical detection^[12–15]. In recent years, TFBG–SPR sensors have been widely explored for various biochemical and biomedical applications, including protein

detection, antibody–antigen interaction monitoring, and other forms of immunosensing^[16–17]. Recent reviews have further highlighted the rapid development of optical fiber SPR biosensors and the particular advantages of grating-assisted plasmonic platforms for biosensing applications.

Despite these advantages, the sensing performance of fiber-optic SPR biosensors is strongly influenced by the immobilization efficiency and activity of biomolecules on the metal surface^[18]. Conventional gold surfaces often provide limited active binding sites for biomolecules, which may restrict the sensitivity and stability of the biosensor. Therefore, developing effective surface functionalization strategies to enhance biomolecule immobilization has become an important research direction for improving the performance of fiber-optic SPR biosensors^[19].

Polydopamine (PDA), inspired by mussel adhesive proteins, has attracted considerable attention as a versatile surface modification material due to its strong adhesion capability and abundant functional groups. PDA coatings can be easily formed on various substrates via self-polymerization and provide active sites for subsequent biomolecule immobilization. Graphene oxide (GO) has also been widely employed in biosensing applications because of its large specific surface area and rich oxygen-containing functional groups, which facilitate efficient biomolecule adsorption and immobilization^[20–22]. The combination of PDA and GO can therefore provide a robust functional interface with enhanced biomolecular binding capability.

In the present work, the TFBG serves as the laser-fabricated photonic element of the sensing platform. It is inscribed in the fiber core by femtosecond-laser direct writing, which enables localized refractive-index modulation and controllable tilted grating planes without chemical etching or mechanical post-processing. Such a grating structure efficiently couples the core mode to backward cladding modes and therefore provides the optical basis for SPR excitation after Au coating. This connection between femtosecond-laser microfabrication, grating-assisted plasmonic coupling, and biomolecular recognition is central to the proposed biosensing scheme.

In this work, a PDA–GO-functionalized TFBG–SPR biosensor is proposed for rapid and label-free detection of human immunoglobulin G (IgG). Compared with previously reported fiber-optic immunosensors, the proposed approach integrates plasmonic sensing with a biomolecule-friendly PDA–GO functional interface, which improves biomolecule immobilization on the sensing surface. In addition, the use of an intensity-based interrogation scheme simplifies the detection system while maintaining reliable sensing performance. The proposed sensor demonstrates rapid response and stable sensing characteristics, providing a promising platform for optical fiber biochemical sensing.

2. Sensor Design and Operating Principle

The proposed immunosensor is based on a TFBG coated with a thin gold film and subsequently functionalized with

a PDA–GO composite recognition layer, as illustrated in Fig. 1(a).

The tilted grating structure enables efficient coupling between the forward-propagating core mode and a series of backward-propagating cladding modes. After gold deposition, selected cladding modes can satisfy the phase-matching condition with surface plasmon waves (SPWs) supported at the metal/external-medium interface, leading to a pronounced attenuation band in the transmission spectrum.

Biomolecular binding occurring near the metal surface modifies the local refractive index environment surrounding the fiber, thereby perturbing the plasmonic coupling condition and inducing measurable spectral variations. This mechanism forms the basis for biochemical sensing using TFBG–SPR structures.

For a uniform TFBG with grating period Λ and tilt angle θ , the magnitude of the grating vector is given by $K = \frac{2\pi}{\Lambda}$. Due to

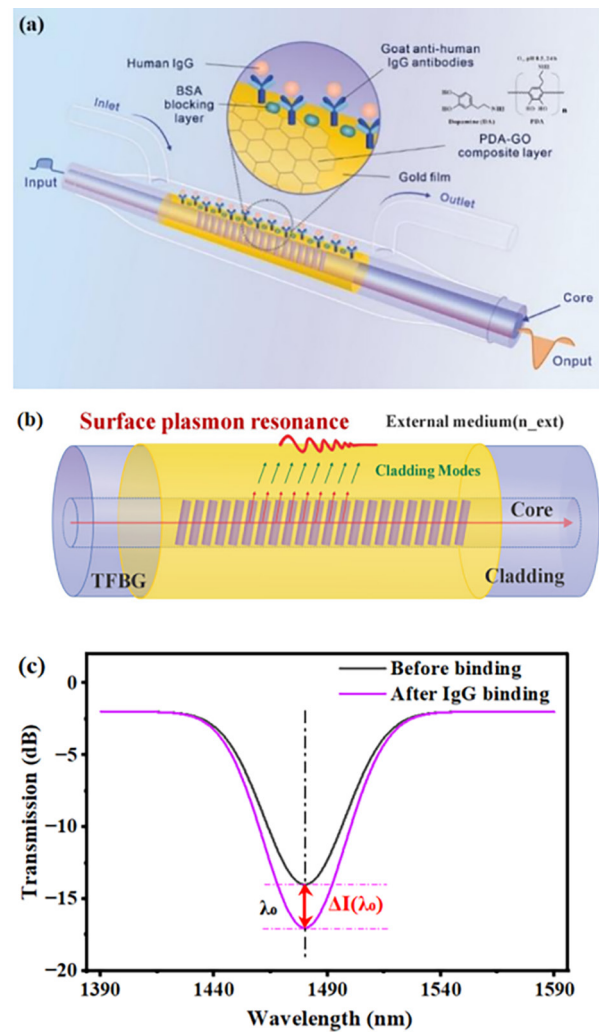


Fig. 1. Schematic diagram of the proposed TFBG-based SPR immunosensor. (a) Functionalization process of the TFBG coated with a gold film and a PDA–GO composite layer for IgG recognition. (b) Surface plasmon resonance coupling mechanism between cladding modes and the metal/external-medium interface. (c) Simulated transmission spectra of the sensor before and after IgG binding.

the grating tilt, the axial component responsible for backward coupling becomes $K_z = \left(\frac{2\pi}{\Lambda}\right)\cos\theta$. The phase matching condition between the forward-propagating core mode and the m -th backward cladding mode can therefore be expressed as

$$\beta_{co}(\lambda) + \beta_{cl,m}(\lambda) = K_z, \quad (1)$$

where β_{co} and $\beta_{cl,m}$ denote the propagation constants of the core mode and the m -th cladding mode, respectively. Using the relation $\beta = \left(\frac{2\pi}{\lambda}\right)n_{eff}$, the resonance wavelength corresponding to the m -th cladding mode can be expressed as

$$\lambda_m = \left(n_{eff,co} + n_{eff,cl,m}\right) \frac{\Lambda}{\cos\theta}. \quad (2)$$

Since the effective refractive index of cladding modes is strongly influenced by the surrounding medium, any modification of the external refractive index or the functional layer deposited on the fiber surface can directly shift the cladding-mode resonances. The tilted structure of the grating breaks the cylindrical symmetry of the fiber and introduces polarization-dependent coupling between the core mode and cladding modes^[8]. As a result, both quasi-TE and quasi-TM cladding modes can be excited by the TFBG structure.

For plasmonic sensing applications, the excitation of surface plasmon waves is predominantly associated with TM-polarized optical fields due to the requirement of an electric-field component normal to the metal surface. The tilted grating structure therefore plays a crucial role in enabling efficient coupling between the cladding modes and the surface plasmon wave supported at the metal-dielectric interface. This polarization-selective coupling significantly enhances the interaction between the guided optical field and the surrounding medium, making TFBG-SPR structures highly suitable for refractive index and biochemical sensing^[11].

When a thin gold film is deposited on the fiber surface, surface plasmon waves can be supported at the metal/external medium interface. The plasmon propagation constant can be expressed as

$$\beta_{sp}(\lambda) = k_0 \sqrt{\frac{\epsilon_m(\lambda)\epsilon_d}{\epsilon_m(\lambda) + \epsilon_d}}, \quad (3)$$

where $k_0 = \frac{2\pi}{\lambda}$ is the free-space wave number, $\epsilon_m(\lambda)$ is the complex permittivity of the gold film, and $\epsilon_d = n_{ext}^2$ represents the permittivity of the external medium. Surface plasmon resonance occurs when the propagation constant of a cladding mode matches the real part of the plasmon propagation constant:

$$\beta_{cl,m}(\lambda) \approx \Re\{\beta_{sp}(\lambda)\}. \quad (4)$$

Under this phase-matching condition, optical power transfers from the guided cladding mode into the lossy plasmonic mode, producing a pronounced attenuation dip in the transmission spectrum. Variations in the external refractive index increase $\Re\{\beta_{sp}(\lambda)\}$, shifting the phase-matching condition

toward longer wavelengths and resulting in a red shift of the SPR resonance. This effect forms the basis for refractive index sensing.

The simulated transmission spectra before and after IgG binding are shown in Fig. 1(c), where a pronounced attenuation dip associated with SPR excitation can be observed. In practical biosensing applications, instead of tracking wavelength shifts, a fixed-wavelength intensity interrogation scheme is adopted to simplify signal processing and improve system robustness. Although wavelength interrogation is widely used in SPR sensors, the wavelength shift induced by a thin biomolecular layer can be relatively small in a TFBG-SPR structure. In contrast, the transmission intensity within the SPR attenuation band is strongly affected by changes in plasmonic coupling strength. Therefore, a fixed-wavelength intensity interrogation scheme was adopted in this work to provide a simple and direct readout without complex peak-tracking procedures. As illustrated in Fig. 1(c), the interrogation wavelength λ_0 is selected near the steep slope region of the SPR attenuation band to maximize sensitivity. The corresponding intensity variation is defined as

$$\Delta I(\lambda_0) = 10 \lg \left(\frac{I_{after}(\lambda_0)}{I_{before}(\lambda_0)} \right), \quad (5)$$

where $I_{before}(\lambda_0)$ and $I_{after}(\lambda_0)$ represent the transmitted optical intensities before and after IgG binding, respectively. ΔI is expressed in dB as the relative intensity variation, with positive and negative values indicating an increase and decrease in intensity, respectively.

Locally, the SPR spectral feature can be approximated by a parametric dip function centered at λ_{SPR} with attenuation depth A and linewidth parameter ω :

$$I(\lambda) \approx I_0 - Af \left(\frac{\lambda - \lambda_{SPR}}{\omega} \right). \quad (6)$$

Biomolecular binding induces perturbations in both the resonance wavelength and the attenuation depth. The first-order intensity variation at the fixed interrogation wavelength can therefore be written as

$$\Delta I(\lambda_0) \approx \frac{\partial I}{\partial \lambda_{SPR}} \delta \lambda_{SPR} + \frac{\partial I}{\partial A} \delta A. \quad (7)$$

This expression indicates that the intensity sensitivity is governed by the local spectral slope of the SPR resonance and the strength of plasmonic coupling. Selecting λ_0 near the maximum slope region enhances the response to small refractive-index perturbations.

In the proposed immunosensor, goat anti-human IgG is immobilized on the PDA-GO-modified gold surface. Upon introduction of human IgG, antigen-antibody binding increases the effective refractive index within the evanescent-field region of the surface plasmon wave. This perturbation modifies the plasmon propagation constant in Eq. (3), thereby altering the phase-matching condition in Eq. (4). Consequently, measurable changes in the SPR attenuation and transmitted intensity at λ_0 are produced, enabling quantitative biosensing.

Overall, the sensing process can be summarized as core-mode excitation → TFBG-induced cladding-mode coupling → SPR excitation → refractive-index perturbation → intensity variation.

3. Sensor Fabrication and Functionalization

3.1. Sensor fabrication

Based on the sensing mechanism described above, the proposed TFBG–SPR biosensor was fabricated by integrating a tilted fiber Bragg grating structure with a plasmonic metal layer and a biochemical recognition interface. The tilted fiber Bragg grating used in this work was inscribed in a standard single-mode fiber by femtosecond-laser direct writing. A femtosecond laser with a central wavelength of 515 nm, a pulse duration of approximately 290 fs, and a repetition rate of 100 kHz was employed as the writing source. The laser beam was focused into the fiber core through a high-numerical-aperture objective, and the fiber was mounted on a precision translation stage together with a rotation stage to control the orientation of the grating planes. During inscription, the average laser power was set to approximately 1.14 mW, corresponding to a pulse energy of about 11.4 nJ. The grating was written by translating the fiber along its axial direction, with a designed grating tilt angle of approximately 10° . In this work, the tilt angle is defined as the angle between the grating plane and the plane perpendicular to the fiber axis. The effective grating length was approximately 12 mm. The femtosecond-laser-induced refractive-index modulation formed periodically tilted grating planes in the core region, enabling efficient coupling from the forward-propagating core mode to backward-propagating cladding modes. Such strong core-to-cladding mode coupling is essential for efficient excitation of SPR when the fiber surface is coated with a thin metal film.

Figure 2(a) shows the microscopic image of the fabricated TFBG. The periodic tilted grating planes can be clearly observed within the fiber core region. The effective grating length is approximately 12 mm, while the fiber core diameter is about $8.9\ \mu\text{m}$, which is consistent with the geometry of standard single-mode fibers. The measured tilt angle of the grating planes is approximately 10° , confirming the successful inscription of the tilted grating structure.

After grating inscription, a thin gold film was deposited onto the fiber surface using a sputtering process. The gold layer enables the excitation of surface plasmon waves at the metal–dielectric interface. When the phase-matching condition between the cladding modes and the surface plasmon wave is satisfied, a pronounced attenuation band appears in the transmission spectrum.

Figures 2(b) and 2(c) show the transmission spectra of the bare TFBG and the Au-coated TFBG, respectively. Compared with the spectrum of the bare TFBG, a distinct attenuation band associated with SPR excitation appears after gold deposition. This SPR-induced attenuation band indicates efficient coupling

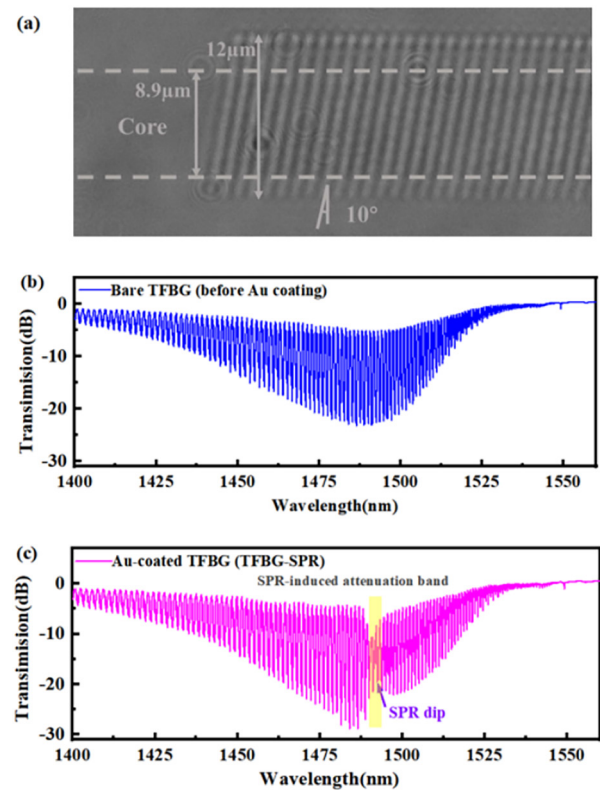


Fig. 2. Fabrication and spectral characteristics of the TFBG–SPR sensor. (a) Microscopic image of the fabricated TFBG. (b) Transmission spectrum of the bare TFBG. (c) Transmission spectrum of the Au-coated TFBG showing the SPR attenuation band.

between the cladding modes and surface plasmon waves supported at the metal–external-medium interface. The resonance dip within this attenuation band exhibits strong sensitivity to variations in the surrounding refractive index, providing the physical basis for biochemical sensing. To further evaluate the sensing capability of the fabricated device, the intrinsic refractive index response of the TFBG–SPR structure was first characterized before biochemical functionalization.

3.2. Intrinsic refractive index performance

Before biochemical functionalization, the intrinsic refractive index (RI) sensitivity of the TFBG–SPR sensor was characterized to evaluate its sensing capability. The surrounding refractive index was varied from 1.300 to 1.350, and the corresponding transmission spectra were recorded.

As shown in Fig. 3, the SPR attenuation band shifts toward longer wavelengths as the external refractive index increases. This spectral evolution originates from changes in the plasmon propagation constant at the metal–dielectric interface. An increase in the external refractive index modifies the phase-matching condition between the cladding modes and the surface plasmon wave, resulting in a red shift of the SPR resonance wavelength. By tracking the minimum of the SPR

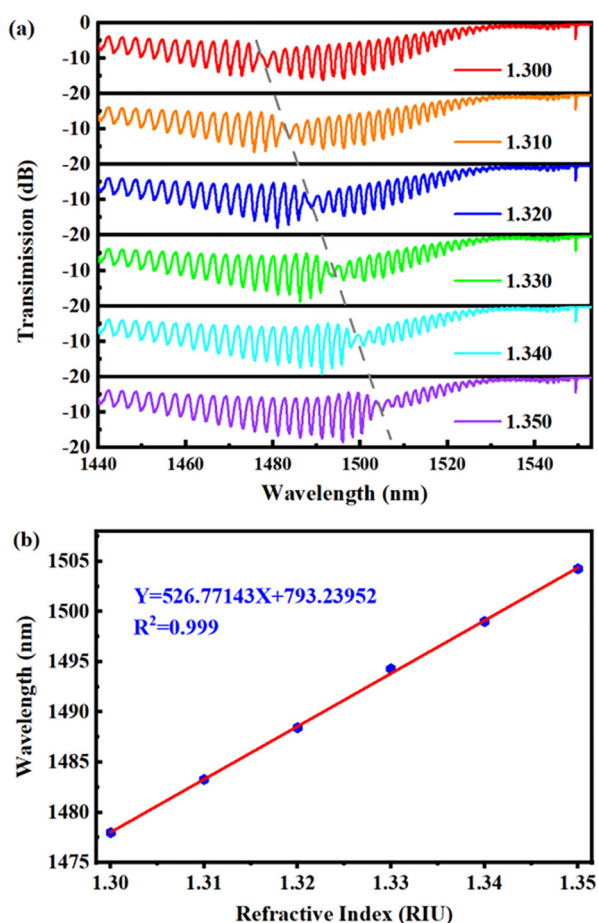


Fig. 3. Intrinsic refractive index sensing performance of the TFBG-SPR sensor. (a) Transmission spectra of the sensor under different surrounding refractive indices ranging from 1.300 to 1.350. A progressive red shift of the SPR attenuation band can be observed as the refractive index increases. (b) Linear relationship between the SPR resonance wavelength and the refractive index, yielding a sensitivity of 526.77 nm/RIU.

attenuation dip, the resonance wavelength variation can be extracted. A linear relationship between the SPR resonance wavelength and the refractive index is observed, yielding a sensitivity of approximately $S = 526.77$ nm/RIU. These results confirm that the TFBG-SPR structure possesses high refractive index sensitivity, providing a reliable sensing platform

for subsequent biochemical functionalization and biosensing experiments.

3.3. Surface functionalization

To enable selective detection of human IgG, the Au-coated TFBG-SPR sensor was modified with a PDA-GO-based biofunctional interface before antibody immobilization, as illustrated in Fig. 4. Briefly, the Au-coated sensing region was first cleaned with freshly prepared piranha solution ($\text{H}_2\text{SO}_4/\text{H}_2\text{O}_2 = 7:3$, v/v) for 30 min, followed by rinsing with ethanol and deionized water. The cleaned Au surface was then modified with a $1 \text{ mg}\cdot\text{mL}^{-1}$ 11-mercaptopundecanoic acid solution to introduce carboxyl groups. The PDA-GO composite was prepared by mixing well-dispersed GO with dopamine hydrochloride in Tris-HCl buffer (10 mmol/L, pH 8.5), where dopamine underwent oxidative self-polymerization under mild alkaline conditions. The obtained PDA-GO composite was subsequently introduced onto the modified Au surface to provide abundant functional groups and a larger effective surface area for biomolecule immobilization.

After PDA-GO modification, the sensing surface was activated with an EDC/NHS solution to promote covalent coupling between the carboxyl groups on the functional layer and the amino groups of goat anti-human IgG. The antibody solution was then introduced to construct the recognition layer, followed by BSA blocking to reduce nonspecific adsorption. After rinsing with PBS, the functionalized TFBG-SPR sensor was used for human IgG detection. Through this process, the PDA-GO-assisted interface enabled stable antibody immobilization and specific capture of human IgG near the Au surface, leading to a local refractive-index change and a measurable SPR intensity response.

Figure 4(b) shows the temporal evolution of the transmission intensity during the functionalization process. The formation of successive biological layers on the gold surface modifies the effective refractive index at the metal-external-medium interface, leading to measurable variations in the SPR attenuation band. When the sensor is exposed to human IgG solutions with different concentrations, antigen-antibody binding occurs on the sensor surface. This binding process increases the local refractive index within the evanescent-field region of the surface

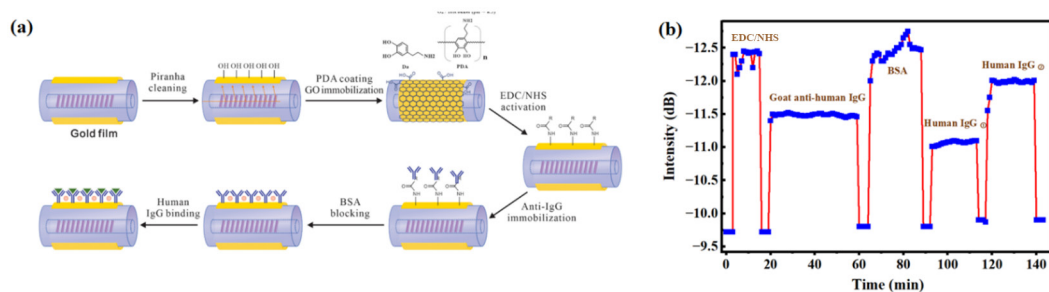


Fig. 4. Surface functionalization process of the TFBG-SPR biosensor. (a) Schematic diagram of the antibody immobilization procedure. (b) Temporal evolution of the transmission intensity during the functionalization.

plasmon wave, thereby perturbing the SPR coupling condition. Instead of tracking the resonance wavelength shift, an intensity-based demodulation scheme is adopted in this work. The transmission intensity of a selected cladding mode within the SPR attenuation band is monitored to quantify the sensing response. This approach simplifies the interrogation system while maintaining a measurable sensing response under the present experimental conditions using the TFBG–SPR platform.

4. Experimental Results and Discussion

4.1. Experimental setup

The experimental setup used for IgG detection with the TFBG–SPR biosensor is shown in Fig. 5. A broadband light source (BBS) was used to illuminate the TFBG sensor through a standard single-mode fiber. The transmitted optical signal was monitored using an optical spectrum analyzer (OSA) with a wavelength resolution of 0.02 nm. To enable controlled delivery of analyte solutions, the gold-coated and biofunctionalized TFBG sensor was integrated into a microfluidic channel connected to a syringe pump. The pump was used to precisely inject IgG solutions with different concentrations into the sensing region. During the sensing experiments, IgG solutions were sequentially introduced into the microfluidic system while the transmission spectra of the sensor were continuously recorded in real time. All measurements were performed at room temperature to maintain experimental stability. Because the core mode of the TFBG is primarily sensitive to temperature rather than the surrounding refractive index, it was employed as an internal reference to compensate for possible temperature fluctuations during the sensing process.

The sensing mechanism relies on monitoring the intensity variation of a selected cladding mode located within the SPR attenuation band. When antigen–antibody binding occurs on the gold surface, the local refractive index near the metal interface changes, thereby modifying the coupling condition between the cladding modes and the surface plasmon wave. As a result, the transmission intensity of the selected cladding mode varies accordingly, enabling quantitative detection of IgG.

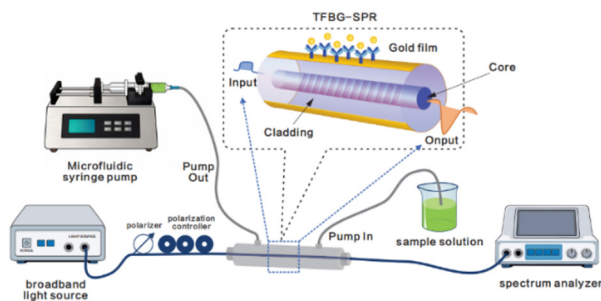


Fig. 5. Experimental setup of the TFBG–SPR biosensor for IgG detection. The inset shows the structure of the Au-coated TFBG–SPR sensor integrated with the microfluidic system.

4.2. IgG detection

The sensing performance of the functionalized TFBG–SPR sensor was evaluated through the immunoreaction between goat anti-human IgG antibodies and human IgG molecules.

Figure 6 presents the transmission spectra of the sensor before and after antigen–antibody binding. After the introduction of the human IgG solution, the transmission intensity of the selected cladding mode located within the SPR attenuation band decreases significantly, while the resonance wavelength exhibits only a negligible shift. The observed intensity decrease is caused by the specific binding of human IgG to the immobilized goat anti-human IgG antibodies on the functionalized Au surface. The formation of antigen–antibody complexes increases the local refractive index near the Au surface, which perturbs the SPR phase-matching condition and changes the coupling strength between the cladding mode and the surface plasmon mode. Consequently, more optical power is transferred to the lossy plasmonic mode, leading to a decrease in the transmitted intensity of the selected cladding mode.

This behavior originates from the relatively small refractive index variation induced by biomolecular binding on the gold surface. Instead of producing a pronounced wavelength shift, the antigen–antibody interaction mainly perturbs the coupling strength between the cladding modes and the surface plasmon wave at the metal–external-medium interface.

Consequently, the spectral feature within the SPR attenuation band becomes highly sensitive to the local refractive index variation near the metal surface. In this case, monitoring the intensity variation of the cladding mode provides a more effective sensing strategy than tracking the resonance wavelength shift.

Based on this principle, an intensity-based demodulation scheme was adopted in this work. A specific cladding mode within the SPR attenuation band was selected, and its transmission intensity variation was used as the sensing parameter for IgG detection.

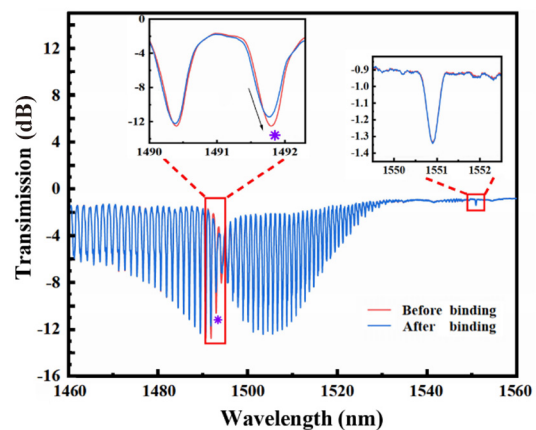


Fig. 6. Transmission spectra of the TFBG–SPR biosensor before and after human IgG binding. The left inset highlights the SPR-sensitive cladding mode used for intensity interrogation, while the right inset shows a reference cladding mode with negligible spectral variation.

As shown in Fig. 6, the left inset highlights the SPR-sensitive cladding mode used for intensity interrogation, while the right inset shows a reference mode that remains nearly unchanged after IgG binding, confirming that the observed intensity variation originates from the SPR interaction rather than external disturbances.

4.3. Quantitative analysis

The quantitative sensing performance of the proposed TFBG–SPR biosensor was further evaluated by introducing human IgG solutions with concentrations of 20, 30, 40, 50, and 60 $\mu\text{g}\cdot\text{mL}^{-1}$.

Figure 7(a) presents the enlarged transmission spectra of the selected SPR-sensitive cladding mode at different IgG concentrations. As the IgG concentration increases, the transmission intensity of the cladding mode within the SPR attenuation band gradually decreases. This behavior results from the increasing amount of antigen–antibody complexes formed on the gold

surface, which modifies the local refractive index in the evanescent-field region and consequently perturbs the plasmonic coupling condition.

To quantitatively analyze the sensing response, the intensity variation of the selected cladding mode was extracted and plotted as a function of IgG concentration, as shown in Fig. 7(b). The experimental data exhibit a clear monotonic decrease in transmission intensity with increasing IgG concentration.

The measured data can be well fitted using an exponential function, yielding a correlation coefficient of $R^2 = 0.997$, which indicates a strong correlation between the intensity variation and IgG concentration. Such a high fitting coefficient confirms the reliability of the proposed sensing scheme for quantitative biochemical detection.

Based on the measured intensity variation over the tested concentration range from 20 to 60 $\mu\text{g}\cdot\text{mL}^{-1}$, the average sensitivity of the proposed sensor is estimated to be approximately $0.0475 \text{ dB } (\mu\text{g}\cdot\text{mL}^{-1})^{-1}$. These results demonstrate that the TFBG–SPR biosensor provides a stable, monotonic, and quantitative response to IgG concentration, indicating its potential for rapid and reliable biochemical sensing applications. The limit of detection (LOD) is estimated as $\text{LOD} = 3\sigma/S$, where S is the fitted sensitivity and σ is the standard deviation of the baseline noise. Here, σ was obtained from the intensity fluctuation of the selected SPR-sensitive cladding mode recorded under blank/PBS conditions before IgG introduction. Using the measured baseline noise and the sensitivity of $0.0475 \text{ dB } (\mu\text{g}\cdot\text{mL}^{-1})^{-1}$, the LOD was calculated to be approximately $0.13 \mu\text{g}\cdot\text{mL}^{-1}$.

4.4. Dynamic response and sensor performance

The dynamic response is qualitatively analyzed to illustrate the sensing behavior. Figure 8 illustrates the temporal response of the sensor when exposed to human IgG solutions with different concentrations. Figure 8(a) shows the spectral evolution of the selected SPR-sensitive cladding mode during the binding process. After the introduction of the analyte solution, the transmission intensity decreases rapidly due to the formation of antigen–antibody complexes on the functionalized gold surface.

As the immunoreaction proceeds, the transmission signal gradually stabilizes and approaches an equilibrium state, indicating that the binding process reaches saturation. The observed spectral evolution confirms that the SPR-sensitive cladding mode responds effectively to the local refractive index variation induced by biomolecular interactions.

The dynamic response curves for different IgG concentrations are presented in Fig. 8(b). A rapid intensity change is observed immediately after the introduction of the analyte solution, followed by a gradual stabilization of the signal. The response time was estimated as the time required for the intensity signal to approach a stable level after the IgG solution was introduced. Based on the dynamic response curves, the signal reached a stable state within approximately 1 min under the present experimental conditions, indicating a relatively fast immunosensing response. The response time of the sensor is

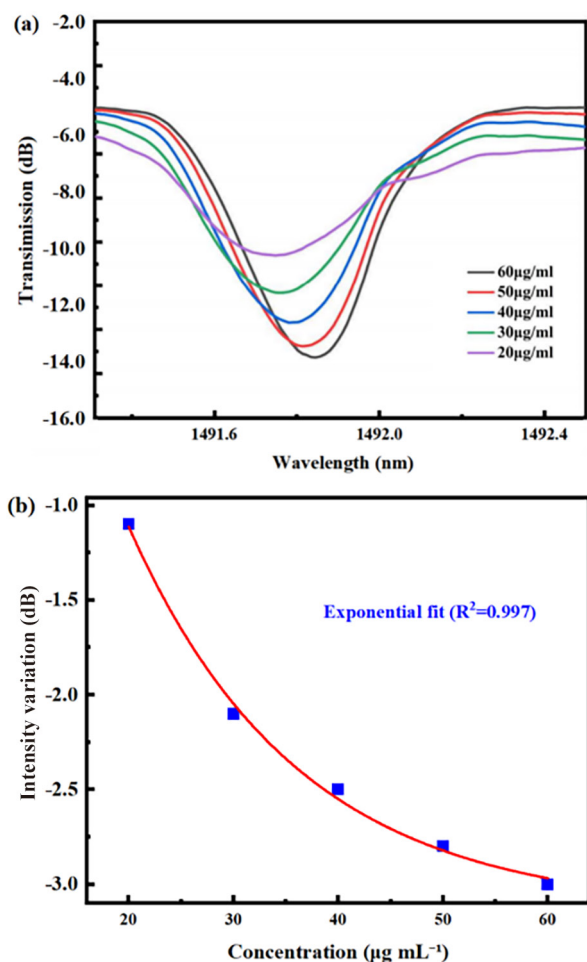


Fig. 7. Concentration-dependent response of the TFBG–SPR biosensor to human IgG. (a) Enlarged transmission spectra of the selected SPR-sensitive cladding mode for different IgG concentrations. (b) Transmission intensity variation of the selected cladding mode as a function of IgG concentration with exponential fitting ($R^2 = 0.997$).

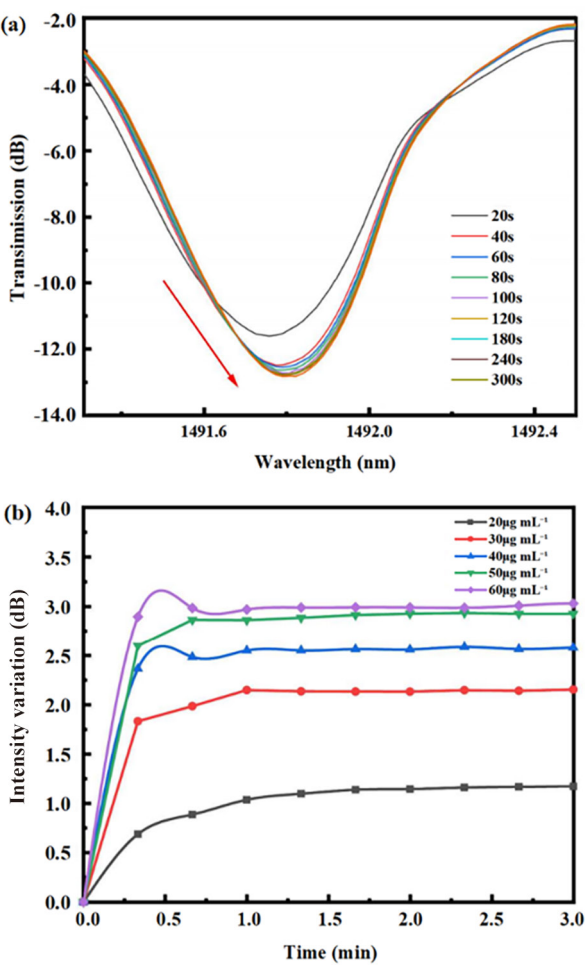


Fig. 8. Dynamic response of the TFBG-SPR biosensor to human IgG. (a) Temporal evolution of the selected SPR-sensitive cladding mode during the binding process. (b) Real-time intensity variation of the selected cladding mode for different IgG concentrations.

within approximately 1 min, demonstrating a relatively fast detection capability. In addition, higher IgG concentrations produce larger transmission intensity variations, which are consistent with the increased amount of antigen-antibody complexes formed on the sensor surface.

To further evaluate the sensing performance, the proposed sensor was compared with previously reported optical fiber biosensors for IgG detection, as summarized in Table 1.

As shown in Table 1, the reported IgG-related optical fiber immunosensors differ in sensing structure, detection method, tested concentration range, and response criterion. The proposed TFBG-SPR biosensor exhibits a clear intensity response in the tested range of 20–60 µg·mL⁻¹, with a sensitivity of 0.0475 dB (µg·mL⁻¹)⁻¹ and an estimated LOD of 0.13 µg·mL⁻¹. Compared with wavelength-tracking schemes, the intensity-based interrogation used in this work provides a simple readout strategy. In addition, the PDA-GO-assisted interface enables a relatively rapid response of approximately 1 min under the present experimental conditions.

5. Conclusion

In this work, a PDA-GO-functionalized TFBG-SPR biosensor was proposed and experimentally demonstrated for label-free detection of IgG. The sensor was fabricated by femtosecond-laser inscription of a TFBG in a standard single-mode fiber, followed by gold coating and PDA-GO surface functionalization to enhance biomolecular immobilization capability.

Compared with conventional wavelength-based SPR fiber sensors, the proposed intensity-modulated approach simplifies the interrogation system and improves robustness for practical biochemical sensing. By employing an intensity-based interrogation scheme based on an SPR-sensitive cladding mode, the proposed sensor enables simplified signal demodulation while maintaining reliable sensing performance. Experimental results demonstrate a clear concentration-dependent response to IgG within the range of 20–60 µg·mL⁻¹, with an exponential fitting coefficient of $R^2 = 0.997$ and an average intensity sensitivity of 0.0475 dB (µg·mL⁻¹)⁻¹. The sensing signal reached a stable level within approximately 1 min under the present experimental conditions. Benefiting from the TFBG-SPR sensing structure and the PDA-GO functional interface, the proposed biosensor provides a compact optical fiber platform for label-free IgG detection. Further optimization and validation with different biomolecular targets will be useful for evaluating its broader applicability in optical fiber biosensing.

Acknowledgements

This work was supported by the National Key Research and Development Program of China (No. 2024YFB3213700)

Table 1. Comparison of Optical Fiber Biosensors for Human IgG Detection.

Detection method	Tested range (µg·mL ⁻¹)	Sensitivity	LOD (µg·mL ⁻¹)	Response time (min)	Ref.
Wavelength shift	0.05–0.5	1.94 nm (µg·mL ⁻¹) ⁻¹	3.4	<3	[23]
Interferometric demodulation	10–50	0.5 dB (µg·mL ⁻¹) ⁻¹	0.028	25	[24]
Reflectivity modulation	0.1–100	Reflectivity change	—	35	[25]
Intensity modulation	20–60	0.0475 dB (µg·mL ⁻¹) ⁻¹	0.13	~1	This work

and the National Natural Science Foundation of China (Nos. T2421003 and U22A2088).

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