

Dual channel planar waveguide surface plasmon resonance biosensor for an aqueous environment

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Abstract: A theoretical study on a dual channel planar waveguide surface plasmon biosensor is presented in this paper. The proposed device consists of a planar waveguide with two active regions. It has been demonstrated that the proposed waveguide sensor can be configured to operate in either multi analyte or self referencing mode. The channel discriminative property of the device is investigated using an eigenmode solver with perfectly matched layers (PML).

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References and links

1. J. Homola, S. Yee, and G. Gauglitz, "Surface plasmon resonance sensors: review," *Sens. Actuators B Chem.* **54**(1-2), 3–15 (1999).
2. C. Mouvet, R. D. Harris, C. Maciag, B. J. Luff, J. S. Wilkinson, J. Piehler, A. Brecht, G. Gauglitz, R. Abuknesha, and G. Ismail, "Determination of simazine in water samples by waveguide surface plasmon resonance," *Anal. Chim. Acta* **338**(1-2), 109–117 (1997).
3. A. Nooke, U. Beck, A. Hertwig, A. Krause, H. Kruger, V. Lohse, D. Negendank, and J. Steinbach, "On the application of gold based SPR sensors for the detection of hazardous gases," *Sens. Actuators B Chem.* **149**(1), 194–198 (2010).
4. J. Homola, J. Dostálek, S. F. Chen, A. Rasooly, S. Jiang, and S. S. Yee, "Spectral surface plasmon resonance biosensor for detection of staphylococcal enterotoxin B in milk," *Int. J. Food Microbiol.* **75**(1-2), 61–69 (2002).
5. V. V. Koubová, E. Bryndaa, L. Karasová, J. Škvor, J. Homola, J. Dostálek, P. Tobiška, and J. Rošický, "Detection of foodborne pathogens using surface plasmon resonance biosensors," *Sens. Actuators B Chem.* **74**, 100–105 (2001).
6. T. T. Goodrich, H. J. Lee, and R. M. Corn, "Direct detection of genomic DNA by enzymatically amplified SPR imaging measurements of RNA microarrays," *J. Am. Chem. Soc.* **126**(13), 4086–4087 (2004).
7. J. Homola, H. Vaisocherová, J. Dostálek, and M. Pilarik, "Multi-analyte surface plasmon resonance biosensing," *Methods* **37**(1), 26–36 (2005).
8. S. Patskovsky, M. Meunier, P. N. Prasad, and A. V. Kabashin, "Self-noise-filtering phase-sensitive surface plasmon resonance biosensing," *Opt. Express* **18**(14), 14353–14358 (2010).
9. X. D. Hoa, A. G. Kirk, and M. Tabrizian, "Towards integrated and sensitive surface plasmon resonance biosensors: a review of recent progress," *Biosens. Bioelectron.* **23**(2), 151–160 (2007).
10. J. Dostálek, J. Ctyroky, J. Homola, E. Brynda, M. Skalský, P. Nekvindová, J. Spirková, J. Škvor, and J. Schröfel, "Surface plasmon resonance biosensor based on integrated optical waveguide," *Sens. Actuators B Chem.* **76**, 8–12 (2001).
11. J. R. Sambles, G. W. Bradbery, and F. Yang, "Optical excitation of surface plasmons: an introduction," *Contemp. Phys.* **32**(3), 173–183 (1991).
12. S. A. Maier, *Plasmonics: Fundamentals and Applications*, Springer, New York, 40 – 52 (2007).
13. C. Jung, S. Yee, and K. Kuhn, "Integrated-Optics Wave-Guide Modulator Based On Surface-Plasmon Resonance," *J. Lightwave Technol.* **12**(10), 1802–1806 (1994).
14. W. Sellmeier, "Zur Erklärung der abnormen Farbenfolge im Spectrum einiger Substanzen," *Ann. Phys.* **219**(6), 272–282 (1871).
15. CAMFR, <http://camfr.sourceforge.net>.
16. COMSOL, <http://www.comsol.com>.
17. P. Yeh, *Optical Waves in Layered Media*, Wiley, New York, (1988).
18. J. Chilwell, and I. Hodgkinson, "Thin-films field-transfer matrix theory of planar multilayer waveguides and reflection from prism-loaded waveguides," *J. Opt. Soc. Am. A* **1**(7), 742–753 (1984).

19. A. Ghatak, K. Thyagarajan, and M. Shenoy, "Numerical analysis of planar optical waveguides using matrix approach," *J. Lightwave Technol.* **5**(5), 660–667 (1987).
20. R. Symms, and J. Cozens, *Optical Guided Waves and Devices*, (McGraw-Hill, London, 1992).
21. P. B. Johnson, and R. W. Christy, "Optical-Constants of Noble-Metals," *Phys. Rev. B* **6**(12), 4370–4379 (1972).
22. M. L. Nesterov, A. V. Kats, and S. K. Turitsyn, "Extremely short-length surface plasmon resonance devices," *Opt. Express* **16**(25), 20227–20240 (2008).
23. R. D. Harris, and J. S. Wilkinson, "Waveguide surface Plasmon resonance sensors," *Sens. Actuators B Chem.* **29**(1-3), 261–267 (1995).
24. E. K. Akowuah, T. Gorman, and S. Haxha, "Design and optimization of a novel surface plasmon resonance biosensor based on Otto configuration," *Opt. Express* **17**(26), 23511–23521 (2009).
25. R. Levy, and S. Ruschin, "Design of a Single-Channel Modal Interferometer Waveguide Sensor," *IEEE Sens. J.* **9**(2), 146–153 (2009).
26. R. Jha, and A. K. Sharma, "High-performance sensor based on surface plasmon resonance with chalcogenide prism and aluminum for detection in infrared," *Opt. Lett.* **34**(6), 749–751 (2009).
27. K. Wang, Z. Zheng, Y. L. Su, Y. M. Wang, Z. Y. Wang, L. S. Song, J. Diamond, and J. S. Zhu, "High-Sensitivity Electro-Optic-Modulated Surface Plasmon Resonance Measurement Using Multilayer Waveguide-Coupled Surface Plasmon Resonance Sensors," *Sens. Lett.* **8**(2), 370–374 (2010).
28. M. Piliarik, L. Párová, and J. Homola, "High-throughput SPR sensor for food safety," *Biosens. Bioelectron.* **24**(5), 1399–1404 (2009).

1. Introduction

Surface plasmon resonance (SPR) sensors provide high sensitivity without the use of molecular labels [1]. SPR biosensors have found extensive application in the analysis of biomolecular interactions (BIA) and detection of chemical and biological analytes [1], where they provide benefits of real-time, sensitive and label-free technology. They have been used for detection of various chemical and biological compounds in areas such as environmental protection [2,3], food safety [4,5] and medical diagnostics [6].

There is a growing interest in the development of SPR sensor platforms capable of discriminating specific sensor response (resulting from capture of analyte molecules by bio molecular recognition) from non – specific response, which is mainly due to temperature fluctuation, analyte composition variation and adsorption of non – target molecules on the sensor surface [7,8]. In addition to sensor discrimination, there is also an interest in sensor platforms capable of simultaneous detection of multiple analytes. These requirements can be met by multichannel sensors containing sensing channels with bio receptors for detection of specific analytes and reference channels responding to non – specific effects [7].

Most multichannel SPR biosensors are generally based on the prism-coupled SPR configuration, which is simple, robust and highly sensitive. However, there are challenges with regards to miniaturization and integration of such structures with existing optical systems [9].

Waveguide based SPR sensor designs can provide highly integrated, multichannel and robust sensing devices as these structures can accommodate several sensing elements on a single platform [9]. The sensing elements can be micro-fabricated in parallel or in series where wavelength division multiplexing techniques may be used to extract the signal from the different sensing elements [9]. This makes it possible to integrate a reference signal which can nullify the effect of environmental changes (temperature changes, noise source) [8–10].

In this paper, we propose a dual channel planar waveguide SPR biosensor based on spectral interrogation. The proposed sensor, as shown in Fig. 1, consists of two channels with different active elements per channel (Gold is used for channel A whilst silver is used for channel B). We intend to demonstrate that the proposed device can be used as a self referencing device to eliminate non specific sensor responses. The multi analyte detection functionality of the proposed device will also be elucidated.

A brief theory regarding waveguide SPR sensors is given in the next section, which will be followed by the discussion of simulation results and finally conclusions in the last section.

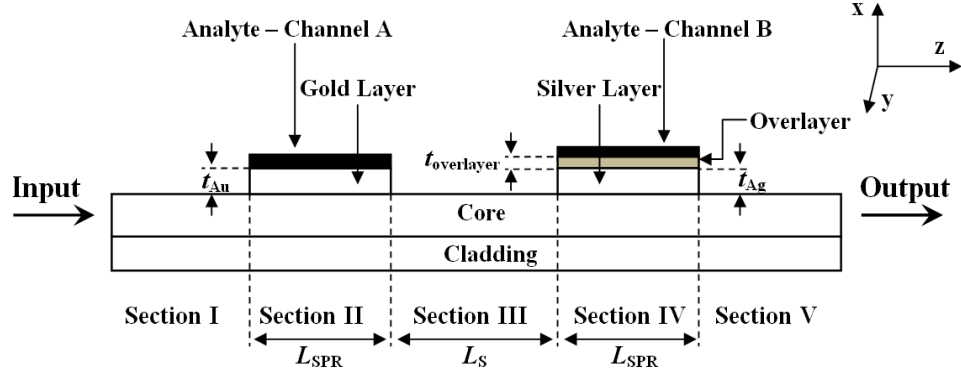


Fig. 1. Cross section of the proposed dual channel waveguide SPR sensor showing various sections.

2. Theory

In this section, we will investigate the nature of a propagating surface plasmon wave (SPW) between the interface of a metal and a dielectric. The implications of having a dielectric and a metal adjacent to each other, with positive and negative relative permittivities (ϵ_r) respectively, will be analysed (Fig. 2). We will only consider TM polarized waves due to the requirement of excitation of SPWs by incident waves [1,11,12].

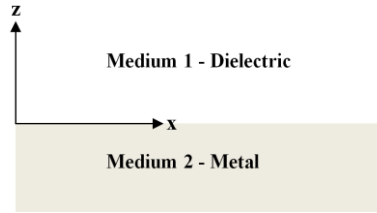


Fig. 2. Geometry for SPW propagation at a single interface between a metal and a dielectric.

If the $x - y$ plane is assumed to be the interface plane, then for a wave propagating in the x direction only, then in medium 1,

$$E_1 = (E_{x1}, 0, E_{z1}) e^{i(k_x x - \omega t)} e^{ik_{z1} z}, \quad (1a)$$

$$H_1 = (0, H_{y1}, 0) e^{i(k_x x - \omega t)} e^{ik_{z1} z}. \quad (1b)$$

Similarly, in medium 2,

$$E_2 = (E_{x2}, 0, E_{z2}) e^{i(k_x x - \omega t)} e^{ik_{z2} z}, \quad (1c)$$

$$H_2 = (0, H_{y2}, 0) e^{i(k_x x - \omega t)} e^{ik_{z2} z}, \quad (1d)$$

where E_i and H_i ($i = 1, 2$) are the electric and magnetic field components respectively, in the two media and $k_i \equiv k_{zi}$, k_{xi} ($i = 1, 2$) is the component of the wave vector perpendicular to the interface in the two media. If we apply Maxwell's equation $\nabla \cdot E = 0$ to Eqs. (1a) and (1c), we obtain;

$$E_{z1} = -E_{x1} \frac{k_x}{k_{z1}}, \quad (2a)$$

$$E_{z2} = -E_{x2} \frac{k_x}{k_{z2}}. \quad (2b)$$

Using Maxwell's equation $\nabla \times E = -\mu \frac{\partial H}{\partial t}$, with $\mu = \mu_0$ results in:

$$H_{y1} = \frac{\omega E_{x1} \varepsilon_1 \varepsilon_0}{k_{z1}}, \quad (3a)$$

$$H_{y2} = \frac{\omega E_{x2} \varepsilon_1 \varepsilon_0}{k_{z2}}. \quad (3b)$$

The requirement of continuity of the tangential E and H at the interface of the media ($z = 0$), implies $H_{y1} = H_{y2}$ and $E_{x1} = E_{x2}$ which gives a relationship between the relative permittivities and the normal components of the wave vectors in both media as:

$$\frac{\varepsilon_1}{k_{z1}} = \frac{\varepsilon_2}{k_{z2}}. \quad (4)$$

In addition, we have:

$$k_{z1} = -i\sqrt{(k_x^2 - \varepsilon_1 k^2)}, \quad \text{which requires } k_x^2 > \varepsilon_1 k^2, \quad (5a)$$

$$k_{z2} = -i\sqrt{(k_x^2 - \varepsilon_2 k^2)}, \quad \text{which requires } k_x^2 > \varepsilon_2 k^2, \quad (5b)$$

where $k = \frac{\omega}{c}$. A requirement for us to have a trapped surface wave with exponential decays into both media is that $ik_{z1} > 0$ whilst $ik_{z2} < 0$. This implies both wave vectors are imaginary with opposite signs, meaning ε_1 and ε_2 are of opposite sign. Now if we substitute Eqs. (5a) and (5b) into Eq. (4), we get:

$$k_x = k \sqrt{\left(\frac{\varepsilon_1 \varepsilon_2}{\varepsilon_1 + \varepsilon_2} \right)}. \quad (6)$$

It can be inferred from Eq. (6) that for k_x to be real, the requirement for a propagating mode, with negative ε_2 (metal), is that $|\varepsilon_2| > \varepsilon_1$. Hence, the trapped surface wave now satisfies Maxwell's equations and boundary conditions with real k_x and appropriate k_z , provided $|\varepsilon_2| > \varepsilon_1$ and $\varepsilon_2 < 0$. The above analysis have assumed real ε values which results in a surface wave having purely real k_x which is larger than $\sqrt{\varepsilon_1} k$, the maximum value for the dielectric medium (medium 1).

However, real metals have resistive scattering properties which lead to damping of oscillations created by the incident electric (E) field [11]. This results in an imaginary component in ε , ε_i . This means for medium 2 (metal), $\varepsilon_2 = \varepsilon_{2r} + i\varepsilon_{2i}$. Equation (6) now becomes:

$$k_x = k \sqrt{\left(\frac{\varepsilon_1 (\varepsilon_{2r} + i\varepsilon_{2i})}{\varepsilon_1 + \varepsilon_{2r} + i\varepsilon_{2i}} \right)}. \quad (7)$$

k_x is now complex and can be represented as $k_x = k_{xr} + ik_{xi}$. If $|k_{xi}| \ll k_{xr}$, with $|\varepsilon_{2r}| \gg \varepsilon_1$ and ε_{2i} , then:

$$k_{xr} \sim k\sqrt{\varepsilon_1} \left(1 - \frac{\varepsilon_1}{2\varepsilon_{2r}}\right), \quad (8a)$$

and

$$k_{xi} = \frac{1}{2}k \frac{\varepsilon_{2i}\sqrt{\varepsilon^3}}{\varepsilon_{2r}^2}. \quad (8b)$$

Hence, the shift in wave vector, Δk_{xr} of this surface plasmon resonance from the critical value, $\sqrt{\varepsilon_1}k$, is given by:

$$\Delta k_{xr} = k_{xr} - \sqrt{\varepsilon_1}k \simeq \frac{1}{2}k \frac{\sqrt{\varepsilon_1^3}}{\varepsilon_{2r}}. \quad (9)$$

It can be seen that the shift is inversely proportional to $|\varepsilon_{2r}|$, while the resonance width is proportional to ε_{2i} and inversely proportional to ε_{2r}^2 . The analysis presented here serves as a tool to optimize material variables of SPR sensors. For example in selecting a metal for a particular design, whilst it may appear beneficial to have a small ε_i , this desire must be balanced with the requirement that we need a large negative value of ε_r . For most metals, while ε_i is generally smallest in the visible part of the spectrum, being larger as we move to infra – red wavelengths, there is an even more rapid increase in $|\varepsilon_r|$. Since the width of the SPR is influenced by $\frac{\varepsilon_i}{\varepsilon_r^2}$ and since ε_r changes faster than ε_i , then the resonance width narrows as the wavelength increases. Metals such as Ag, Au, Al and Cu support a sharp SPR in the visible spectrum [11].

3. Simulation results

The proposed sensor (Fig. 1) has input (I) and output (V) waveguide sections with multilayer (II & IV) sections joined by an intermediate section (III).

The thickness of the core and cladding are fixed at 1 μm and 2 μm respectively, to ensure single mode operation [13]. The core is made of silica which is modeled using the Sellmeier [14] equation;

$$n(\lambda) = \sqrt{1 + \frac{B_1\lambda^2}{\lambda^2 - C_1} + \frac{B_2\lambda^2}{\lambda^2 - C_2} + \frac{B_3\lambda^2}{\lambda^2 - C_3}}, \quad (10)$$

where n is the refractive index, λ , the wavelength in μm , $B_{(i=1,2,3)}$ and $C_{(i=1,2,3)}$ are Sellmeier coefficients. The Sellmeier coefficients used for the core are $B_1 = 0.696166300$, $B_2 = 0.407942600$, $B_3 = 0.897479400$, $C_1 = 4.67914826 \times 10^{-3} \mu\text{m}^2$, $C_2 = 1.35120631 \times 10^{-2} \mu\text{m}^2$ and $C_3 = 97.9340025 \mu\text{m}^2$ [14]. The refractive index of the cladding is assumed to be 1.444.

The length of the active region in both channels A and B is denoted as L_{SPR} whilst the thickness of the gold and silver is given by t_{Au} and t_{Ag} respectively. The two channels are linked by section (III) of length, L_s . A thin overlayer of thickness $t_{\text{overlayer}}$ has been deposited on top of the silver to protect it.

Simulations were carried out using CAMFR, an eigenmode solver with perfectly matched layers (PML) [15]. This ensures the absorption of waves travelling towards the walls, without the introduction of additional reflections, regardless of wavelength, incidence angle or polarisation of the incident wave [15]. The simulation results obtained by CAMFR were verified by benchmarking them against results from a commercial Finite Element software package [16–19]. The relatively high memory requirement and longer computation time of FEM for analysis of the proposed sensor resulted in the choice of CAMFR as the computation tool.

Each section of the sensor is modeled as a multilayer system and CAMFR is used to solve eigenmodes supported therein. The optical power transmitted through a given structure is calculated by analysing the modal coupling between the various sections of the sensor. Sections (I) and (V) are assumed to be identical and lossless. The modal expansion coefficients are calculated using the necessary overlap integrals after the modes in Sections (I) and (II) are located [20]. These modes are then propagated through the length of Section (II) of the sensor before being coupled into Section (V), enabling the power transmitted to be calculated. The permittivity of gold and silver were modelled using data from Johnson and Christy [21].

We begin the analysis by considering a simple case of a single channel waveguide SPR sensor with Gold as the active element.

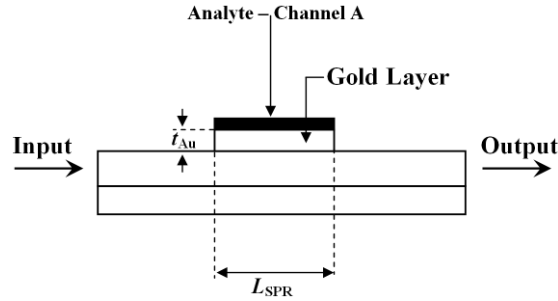


Fig. 3. Cross section of a single channel waveguide SPR sensor optimized for aqueous environment. $t_{Au} = 50\text{nm}$ and $L_{SPR} = 100\mu\text{m}$.

The sensor as shown in Fig. 3, is optimized to operate in an aqueous environment with device parameters of $t_{Au} = 50\text{ nm}$ and $L_{SPR} = 100\text{ }\mu\text{m}$. A broadband (550nm – 850nm) TM polarized light is applied at the input section of the sensor, and the transmission characteristics of the entire system are observed for a change in refractive index of the analyte from 1.33 (aqueous environment) to 1.34. The transmission response is shown in Fig. 4.

Figure 4 illustrates the shift in resonant wavelength for a small change in analyte refractive index of the sensor under consideration. This can be used to determine the sensitivity, S_n of the sensor. The sensitivity (S_n) of an SPR sensor with spectral interrogation is defined as [1]

$$S_n = \frac{\delta\lambda_{res}}{\delta n_{analyte}}, \quad (11)$$

where $\delta n_{analyte}$ is the change in analyte refractive index and $\delta\lambda_{res}$ is the corresponding shift in resonance wavelength. It can be observed from Fig. 4 that there is a shift in resonant wavelength from 609 nm to 624 nm for a change in refractive index of the analyte from 1.33 to 1.34 respectively. By substituting these values into Eq. (11), the sensitivity (S_n) of the sensor can be estimated as 1500 nm/ RIU. Optimisation of material and dimensional variables of planar waveguide SPR sensors has been extensively discussed in [22–25].

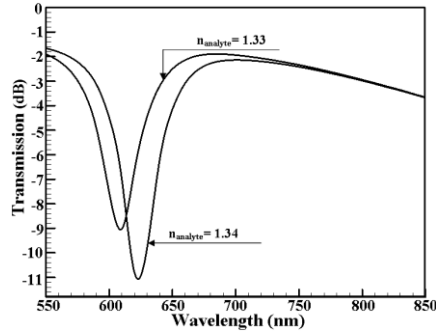


Fig. 4. Transmission properties of a single channel waveguide SPR sensor optimized for aqueous environment. $t_{Au} = 50\text{nm}$ and $L_{SPR} = 100\mu\text{m}$.

The next step in the development of the dual channel sensor is to develop the simple sensor into dual a channel structure shown in Fig. 5. Device parameters used are $t_{Au} = 50\text{ nm}$, $L_S = 10\text{ }\mu\text{m}$ and $L_{SPR} = 45\text{ }\mu\text{m}$.

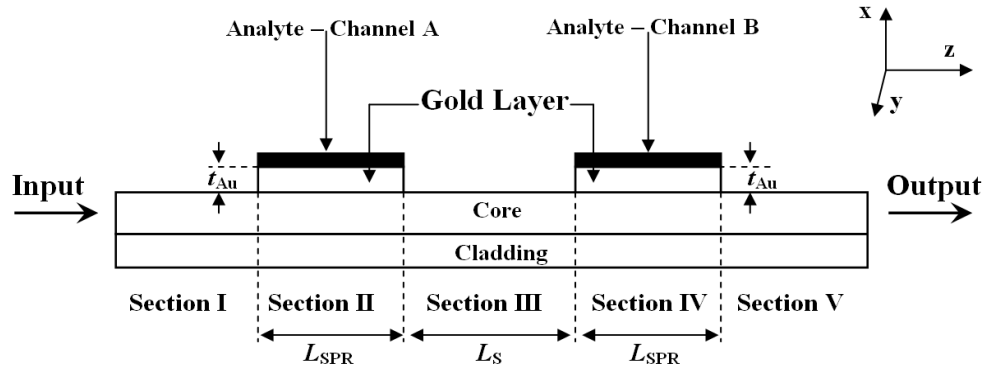


Fig. 5. Cross section of a dual channel waveguide SPR sensor with gold as active material in both channels. $t_{Au} = 50\text{ nm}$, $L_S = 10\text{ }\mu\text{m}$ and $L_{SPR} = 45\text{ }\mu\text{m}$

In order to verify the multi analyte detection capability of the sensor, channel B is immersed in water ($n_{\text{analyte}} = 1.33$) whilst that of Channel A is varied from 1.33 to 1.36. The transmission properties of the sensor are obtained as in the case of the single channel by the use of a broadband TM polarized light wave. Figure 6 illustrates the response of the sensor.

As seen in Fig. 6, some of the curves exhibit two distinct dips corresponding to the resonant wavelengths of the analytes in the respective channels when the refractive index of the analyte in channel A is at least 1.35. It is also evident, the inability of the sensor to discriminate between analytes for the particular scenario of analyte refractive indices of 1.34 and 1.33 in channels A and B respectively. Ideally, the transmission spectrum should show only one dip when identical analytes are put in different channels, whilst discriminating between analytes whenever there is a difference in analyte refractive indices.

In order to solve the problem evident in Fig. 6, the active material in Channel B was changed to silver with a protective overlayer against oxidation. This results in the sensor structure shown in Fig. 1. There are several material and dimensional variables to consider when optimizing the structure for dual channel operation [22–25]. For the sake of simplicity, the thickness of gold (t_{Au}) and silver (t_{Ag}) are fixed at 50nm, whilst L_{SPR} and L_S are kept at 45 μm and 10 μm respectively. The structural variables of interest are $n_{\text{overlayer}}$ – the refractive index of the protective overlayer for silver and $t_{\text{overlayer}}$ – the thickness of the overlayer.

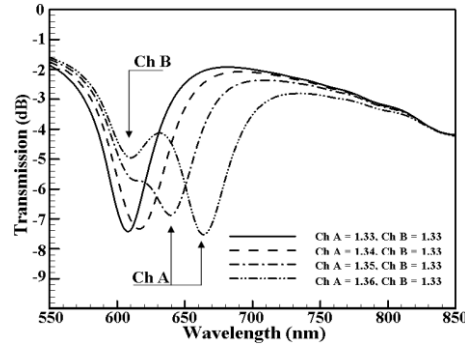


Fig. 6. Transmission properties of a dual channel waveguide SPR sensor with gold as active material in both channels. $t_{Au} = 50$ nm, $L_S = 10$ μ m and $L_{SPR} = 45$ μ m

We begin the analysis by investigating the effect of $n_{overlay}$ on channel discrimination by the SPR spectrum. The analyte refractive indices in channels A and B are fixed at 1.34 and 1.33 respectively. Figure 7 (I and II) illustrate the response of the proposed sensor to a variation in $n_{overlay}$.

It can be observed from Fig. 7(I & II) that for $1.39 \leq n_{overlay} \leq 1.42$, the curves exhibit two distinct dips corresponding to the resonant wavelengths of the respective channels. It also shows that the resonant wavelength in channel B can be tuned to a desired value without affecting that of channel A.

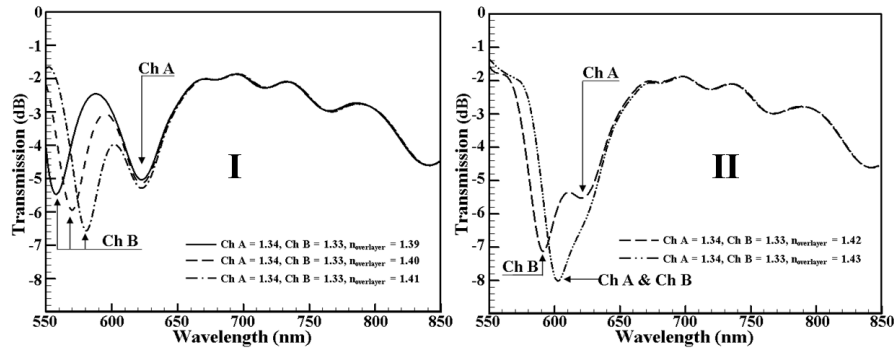


Fig. 7. Transmission properties of a dual channel waveguide SPR sensor with gold and silver as active material in channel A and B respectively. Device parameters: $t_{Au} = 50$ nm, $L_S = 10$ μ m and $L_{SPR} = 45$ μ m, $t_{overlay} = 50$ nm, $n_{overlay}$ is varied from 1.39 to 1.43.

In our next step, we investigate the effect of $t_{overlay}$ on the channel discrimination. This is done by fixing $n_{overlay}$ at 1.40 whilst varying $t_{overlay}$. The results are illustrated in Fig. 8.

Figure 8 shows the dependence of the resonant wavelength in channel B on the thickness of the overlayer. It can be seen to shift from 570 nm to 590 nm for change in $t_{overlay}$ from 50 nm to 70 nm. In addition, the discriminative property of the sensor is seen to deteriorate as $t_{overlay}$ is increased from 50 nm to 70 nm.

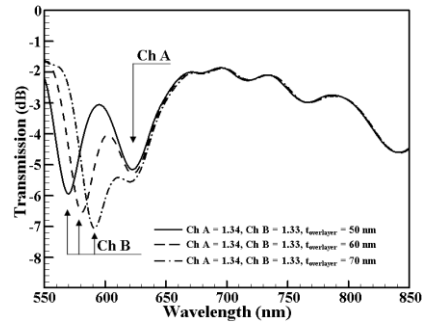


Fig. 8. Transmission properties of a dual channel waveguide SPR sensor with gold and silver as active material in channel A and B respectively. Device parameters: $t_{Au} = 50$ nm, $L_S = 10$ μ m and $L_{SPR} = 45$ μ m, $n_{overlayer} = 1.40$, $t_{overlayer}$ is varied from 50nm – 70nm.

The analysis carried out so far gives a good idea of device parameter tolerance values within which the proposed sensor would maintain its performance. In particular, $1.39 \leq n_{overlayer} \leq 1.43$ for $t_{overlayer} = 50$ nm and $50 \text{ nm} \leq t_{overlayer} \leq 70$ nm for $n_{overlayer} = 1.40$ yield two tolerance criteria for the device. Polymers are prime candidates for the overlayer as they can be easily synthesised for any given refractive index range. In addition, thermo optic polymers can be used for the overlayer in to change the properties of the sensor dynamically. For practical purposes, the length of section III (L_S) needs to be increased for distinguishable sensing between channels. Since section III (L_S) serves as a waveguide for both channels, the significant property affected by the length of L_S is the propagation loss. By performing similar analysis, the maximum acceptable length of L_S is determined as 12 mm [22,23,25].

3.1 Multi Analyte Operation

Multi analyte SPR sensors are capable of simultaneous detection of multiple analytes [7]. In order to configure our proposed structure as a multi analyte SPR sensor, each channel is coated with a functionalizing material to capture the analytes of interest. Device parameters of $t_{Au} = 50$ nm, $t_{Ag} = 50$ nm, $L_S = 10$ μ m, $L_{SPR} = 45$ μ m, $t_{overlayer} = 50$ nm and $n_{overlayer} = 1.40$ are used for the configuration of the proposed structure as a multi analyte SPR sensor.

In order to simulate multi analyte detection capability of the proposed sensor, channels A and B were assumed immersed in analytes of different refractive indices. Figure 9 shows the resulting spectrum. The figure shows that the different analytes have been clearly represented as distinct dips which correspond to the resonant wavelengths of the respective channels. The channels are independent, thus the individual sensitivities can be calculated.

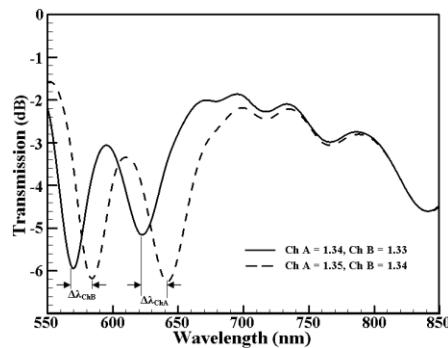


Fig. 9. Transmission properties of the proposed dual channel waveguide SPR sensor configured as a multi analyte detector. Device parameters: $t_{Au} = 50$ nm, $L_S = 10$ μ m and $L_{SPR} = 45$ μ m, $n_{overlayer} = 1.40$, $t_{overlayer} = 50$ nm.

3.2 Self Referencing Operation

Multi channel SPR sensors can be configured as self referencing sensors when some of the channels are coated with bio receptors responding to non – specific effects [7]. The proposed SPR structure can be configured to operate as a self referencing sensor by setting $n_{\text{overlay}} = 1.41$ whilst maintaining the rest of the device parameters as in the case of multi analyte operation.

Channel B of the proposed sensor is immersed in water (1.33) whilst the refractive index of the analyte in channel A is varied from 1.33 to 1.36 to simulate the self referencing capability of the sensor. Figure 10 shows the operation of the sensor with self referencing capability.

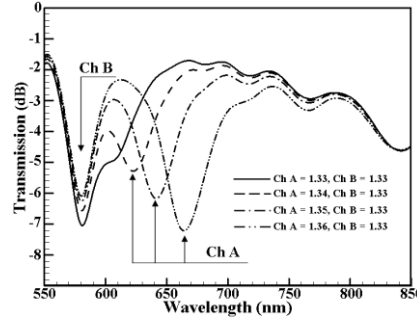


Fig. 10. Transmission properties of the proposed dual channel waveguide SPR sensor configured as a self referencing sensor. Device parameters: $t_{\text{Au}} = 50$ nm, $L_S = 10$ μm and $L_{\text{SPR}} = 45$ μm , $n_{\text{overlay}} = 1.41$, $t_{\text{overlay}} = 50$ nm..

As seen in Fig. 10, the spectrum shows one dip corresponding to the resonant wavelength when the refractive index of the analyte in channel A matches that in B (1.33) and a second dip when the analyte in A is different. The separation between the reference channel (B) and that of A, can be used as a measure of sensitivity. If we denote the separation as λ_{sep} , then the sensitivity of the sensor could also be calculated as

$$S_n = \frac{\lambda_{\text{sep}}}{\delta n_{\text{analyte}}}. \quad (12)$$

Although the resonant wavelength, λ_{res} is susceptible to external influences such as temperature, etc, λ_{sep} remains the same since the change in λ_{res} in both channels will be the same. Hence. Equation (12) becomes more reliable in the presence of external influences.

4. Conclusion

The design and optimization of a dual channel waveguide SPR biosensor has been presented in this paper. It has been demonstrated that by selecting appropriate structural and material parameters, the sensor can be configured to operate as either multi analyte or self referencing sensor.

The multi analyte configuration will be suitable for applications where simultaneous detection of analytes is desired. Self referencing capability enables the sensor to nullify the effect of environmental changes (temperature changes, noise source) [6,8,10]. Since the sensor uses a planar waveguide platform, it is possible to design a new structure which has both multi analyte detection and self referencing abilities whilst keeping the device compact. This can be achieved by micro – fabricating the sensing channels in parallel and series whilst employing wavelength division multiplexing techniques to extract signals from the various channels [9, 10]. With regards to key performance parameters such as sensitivity and FWHM, the proposed device is comparable to the DBR – waveguide and dual LED SPR sensors

reviewed by [9]. However, the proposed structure can incorporate concepts from recent single channel designs [25–28] to realize multi channel sensors with high performance parameters.