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Plasmonic optical sensor for determination of refractive index of human skin tissues

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ABSTRACT

A surface plasmon resonance (SPR) based optical sensor for the determination of refractive index of human skin tissue samples is proposed. Previous experimental results describing variation of refractive index of skin tissue samples with wavelength are considered for theoretical calculations. The angular interrogation method along with glass substrate and Au-Al bimetal layer is considered. The sensor's performance is closely analyzed in terms of well-defined performance indicators in order to achieve reliable and accurate sensing performance. The influence of operating wavelength on the performance of sensor scheme is investigated. Performance comparison for two different substrates (2S2G and SF10) is carried out. The proposed sensor has the potential to provide high sensitivity, precision, and resolution, thereby opening an easy and reliable window for dermatological applications.

1. Introduction

Surface plasmon resonance (SPR) is an extremely sensitive and reliable phenomenon for chemical and biochemical sensing [1–5]. In SPR, a TM (transverse magnetic) polarized light causes the excitation of an electron density oscillation at the metal-dielectric interface. This collective oscillation is known as surface plasmon wave (SPW). A resonance takes place when the energy as well as momentum of both the incident (TM-polarized or p-polarized) wave and SPW coincide. The Kretschmann–Reather attenuated total reflection (ATR) configuration (Fig. 1) is most frequently used in SPR phenomenon, in which a thin metal layer is directly deposited on the base of a light-coupling substrate [6]. The plasmon resonance condition is expressed as:

$$K_0 n_c \sin\theta = K_0 \left(\frac{\varepsilon_m \varepsilon_s}{\varepsilon_m + \varepsilon_s}\right)^{1/2}; K_0 = \frac{2\pi}{\lambda} \quad (1)$$

The term on left hand side is the propagation constant (K_{inc}) of the evanescent wave generated due to ATR of the light (of wavelength λ) incident at an angle θ through the light coupling substrate of refractive index (*RI*) n_c. The right hand term is taken as the real part of complex propagation constant (K_{SPW}) of SPW with ε_m and ε_s being, respectively, the dielectric constants of metal and sensing (dielectric) medium. When the above condition is fulfilled, the resonance appears in the form of a sharp dip of reflected output signal at an angle θ_{SPR} due to strong optical absorption by SPW (Fig. 1). Any small change in n_s ($= \sqrt{\varepsilon_s}$) near the metal-dielectric interface causes a shift in the resonance curve making SPR a very effective and fast-responding sensing technique. Usually, SPR-based measurements are performed with angular interrogation in which λ is kept fixed (i.e., a monochromatic light source is used) and θ is varied.

Apart from response time, which is an experimental characteristic, the performance of SPR sensor is broadly determined in terms of three criteria. First, the shift in resonance angle (i.e., $\delta\theta_{SPR}$) for a given change (δn_s) in n_s should be as large as possible. Second, the full width at half maximum (*FWHM or* $\delta\theta_{0.5}$) corresponding to SPR curves should be as small as possible for precise measurements. Third, the resolution of measurement should be as small as possible so that even the negligibly small changes may be determined.

There are a number of biological and biomedical parameters, where SPR-based biosensing has been proposed or utilized for their detection, such as feline calicivirus [3], antibodies [5], membrane proteins [7], immunoassays [8], human blood-groups [9], Avian influenza virus H5N1 [10], estrone [11], and hemoglobin concentration [12]. However, there are several other biological processes where SPR is yet to be explored. One of such processes may be the determination of *RI* of human skin tissues, which is a key process in establishing the skin's optical response for dermatological applications. Different optical techniques are in use to

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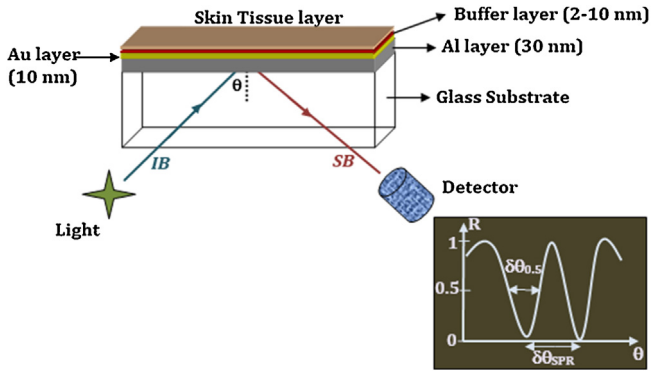

Fig. 1. Proposed SPR sensor probe set-up for the determination of refractive index of human skin tissues. *IB* stands for incoming light beam and *SB* stands for sensed light beam.

monitor different skin properties such as moisture (corneometer), pH, colour (colorimeter), temperature, and viscoelasticity (cutometer) etc. However, the determination of *RI* of skin tissues is an area that is comparatively unexplored. Worth-mentioning is that a precise measurement of RI of human skin tissues is highly relevant for an accomplished skin characterisation (e.g., testing of medication of neurodermatitis patients, reactions of baby skin to irritant alkaline washing agents, and testing of allergy reactions etc.). Among several procedures, optical coherence tomography (OCT) has been tipped as one of the reliable techniques for determination of human skin tissues [13].

The most common challenge in determining the *RI* of the human skin tissues lies in their turbid nature, which leads to an undesired scattering of light upon penetration into the biological samples of human skin tissues. In this context, Ding et al. presented a method of determining the *RI* of the turbid samples of human skin tissue by measuring the angular variation of reflectance [14]. The same method was further extended to find out the imaginary part of the *RI* for both dermis and epidermis tissues [15]. The above work carried out the *RI* measurement for multiple wavelengths ranging between 300 nm (near UV) and 1600 nm (near mid IR). They reported their measurements to be showing the least deviation with the actual value only for selected wavelengths, indicating towards the spectral sensitivity of *RI* measurements. They also reported the spectral variation of *RI*'s real part fitting well with different dispersion relations.

However, minimization of the error and uncertainty associated with *RI* measurement remains an issue to be looked into. The spectral dependence of the *RI* measurements must be negated by getting to a range of operating wavelength that can be used to carry out the measurement for maximum samples. There has to be found out a more efficient method (e.g., SPR) for determining the *RI* of human skin tissues that apart from providing errorless measurements can also optimize the selection of wavelength and other important parameters.

The present work reports the design considerations to enable SPR-based determination of *RI* of human skin tissues by making use of the experimental data provided by Ding et al. The performance of SPR biosensor has been analyzed by taking two different light-coupling substrates (SF10 silica and 2S2G chalcogenide glasses). Previous study shows that Aluminum (Al)-Gold (Au) bimetallic combination can provide much better SPR sensing credentials and stability together compared to the single Al or single Au layer-based designs [16]. Therefore, an Al-Au bimetallic layer has been considered for the present SPR biosensor with angular interrogation. The influence of operating wavelength and the suitability of substrate are studied on proposed sensor's performance in order to identify the appropriate working conditions leading to a highly accurate and reliable SPR-based optical determination of *RI* of human skin tissues.

2. Design considerations

In this section, the modalities of different constituents of the proposed sensor design are systematically discussed along with their physical, biological, and optical properties.

2.1. Light coupling substrate

The light coupling device has to be chosen keeping the operating wavelength in mind. For wavelengths lying in visible range, the glass substrates such as SF10 can be considered, whose wavelength-dependent *RI* (n_c) can be represented in terms of Cauchy's relationship as following expression [17]:

$$n_c(\lambda) = 1.7280 + \frac{0.01342}{\lambda^2} \tag{2}$$

For higher wavelengths lying in infrared range, the chalcogenide glasses (e.g., $Ge_{20}Ga_5Sb_{10}S_{65}$ called 2S2G) have advantage over conventional glasses owing to their higher thermal and chemical stability. The *RI* (n_c) of 2S2G glass is given by following empirical expression [18]:

$$n_c(\lambda) = 2.24047 + \frac{0.02693}{\lambda^2} + \frac{0.00808}{\lambda^4} \tag{3}$$

In Eqs. (2) and (3), λ denotes the operating wavelength (in μm) of the p-polarized light.

2.2. Metal layer

As shown in Fig. 1, the top surface of silica substrate is coated with thin Al layer of thickness d_1 (30 nm in the present scheme). Although Al layer provides the sharpest SPR curve compared with Au, Ag, and Cu [19], it is more prone to oxidation problem than above three SPR-active metals. For this reason, Al layer has to be protected against oxidation that can be done by coating an Au layer of thickness d_2 (much less than d_1) as Au is considered to be not only the most stable SPR-active metal but also provides the largest shift in SPR curves among all SPR-active metals. Such Al-Au bimetallic combination is preferable as it provides way better sensitivity compared with single Au layer [16]. The thickness of Au layer has been taken as 10 nm in the present study. Thus, the total thickness of this Al-Au bimetallic layer is 40 nm. Further, according to free-electron Drude model, the wavelength-dependent complex dielectric function (ε_m) of a metal can be written as:

$$\varepsilon_m(\lambda) = \varepsilon_{mr} + i\varepsilon_{mi} = 1 - \frac{\lambda^2\lambda_c}{\lambda_p^2(\lambda_c + i\lambda)} \tag{4}$$

In above expression, λ_p stands for plasma wavelength, and λ_c stands for collision (or damping) wavelength. For Au, the standard value of λ_p is 168.26 nm and of λ_c is 8.93 μm and for Al, λ_p is 106.57 nm and λ_c is 24.511 μm [19]. However, these values of λ_p and λ_c differ significantly at IR wavelengths and have to be taken accordingly [20].

2.3. Buffer layer

The Au layer in Fig. 1 is followed by a buffer layer of thickness in the vicinity of 1–15 nm. Preferably, this buffer layer should be in the form of a biochemical layer due to two important reasons. First, it prevents the blood sample from being in direct contact with the Au layer which may contaminate the skin tissue sample thereby affecting sensor's performance. Second, the structural compatibility of such a biochemical layer with the skin tissue sample is another