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# Simulation and performance evaluation of fiber optic sensor for detection of hepatic malignancies in human liver tissues

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ABSTRACT

A fiber optic sensor is proposed for the identification of healthy and cancerous liver tissues through determination of their corresponding refractive index values. Existing experimental results describing variation of complex refractive index of liver tissues in near infrared (NIR) spectral region are considered for theoretical calculations. The intensity interrogation method with chalcogenide fiber is considered. The sensor's performance is closely analyzed in terms of its sensitivity at multiple operating wavelengths falling in NIR region. Operating at shorter NIR wavelengths leads to greater sensitivity. The effect of design parameters (sensing region length and fiber core diameter), different launching conditions, and fiber glass materials on sensor's performance is examined. The proposed sensor has the potential to provide high sensitivity of liver tissue detection.



## 1. Introduction

Infrared (IR) spectroscopy technique is extensively used in order to know the physicochemical properties of gasses, liquids, and solids with the help of vibrational motion characteristics of ions or molecules [1]. Most chemical and biological species have their vibrational signatures/absorption domain in IR region, therefore, IR spectroscopy technique is ideal for detection of a wide range of chemicals, pollutants, and biological pathogens [2,3]. IR spectroscopy is a non-destructive analyzing technique, which uses only a small quantity of analyte without any prior preparation. Over the last few decades, fiber-optic evanescent wave spectroscopy (FEWS) has become a popular analytical technique [4,5]. Use of Evanescent field makes it possible to obtain qualitative and quantitative information about the analyte properties (composition, concentration etc). The basic working principle for evanescent wave optical fiber sensor is based on attenuated total internal reflection (ATR) [6,7]. Ample attention has been drawn to IR spectroscopy technique with evanescent wave optical fibers on account of their widespread applications in bio-sensing, continuous monitoring of concentrations of reactants in chemical processes and in the study of the absorption spectra of liquids and pastes [5,8,9]. Fundamentally, FEWS based technique is dependent upon the absorption of the evanescent wave which evolves on the fiber surface. When an electromagnetic wave (EMW) propagates inside a fiber core, an exponentially decaying portion of this wave, called the evanescent wave, extends outside the core region and interacts with the medium in contact as shown in Fig. 1 [10].

Absorption of EMW depends upon the imaginary part of contact medium's complex refractive index ($n^*$) as given below [11]:

$$\alpha = \frac{4\pi n^*}{\lambda} \tag{1}$$

In Eq. (1), $\alpha$ is the absorption coefficient of the analyzing medium and $\lambda$ denotes the wavelength of EMW. For an optical fiber, the power transmitted in the presence of absorbing fluid is given as [12]:

$$P = P_o \exp(-\gamma L/D) \tag{2}$$

In above equation, $P_0$ is the output power when the cladding is not removed (reference power), $\gamma$ is absorption coefficient of the evanescent wave, $L$ is length of the unclad region (sensing length) and D represents core diameter.

Chalcogenide glass-based optical fibers have attracted attention in the development of optical devices due to their low loss transmission characteristics in the mid to far IR wavelength regions from 1 μm to nearly 20 μm [13]. It is a well-known fact that the visible radiation can cause photo-damage or photo-toxicity to biosamples but this is not the case with IR radiation [14,15]. Due

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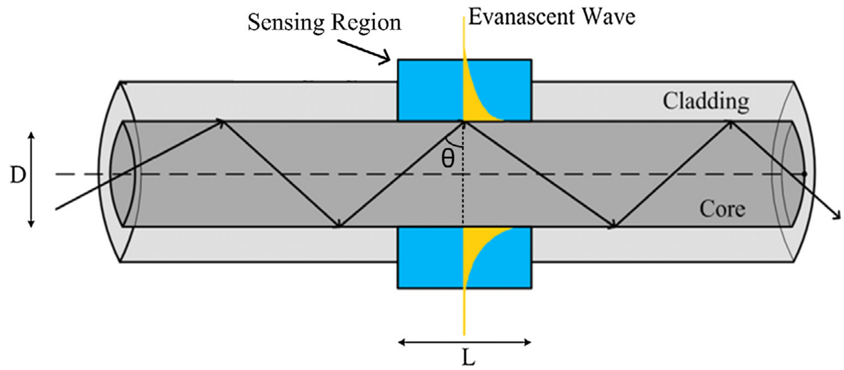


**Fig. 1.** Schematic diagram of evanescent wave based fiber sensor. Some portion of cladding is removed for better absorption of evanescent wave by the sensing region (of length L). Fiber core diameter is D.

to this reason, the chalcogenide fiber based sensors operated in IR can be suitable for detection of various biosamples [16,17]. One of the possible applications can be for detection of cancerous tissues of different body parts *e.g.*, liver. The logic behind this is that the healthy and cancerous tissue samples must have different dispersion relationship due to corresponding variations in their chemical and biological compositions. Human body comprises of spatially distributed tissues with distinct properties [18]. The conduction in any biological tissue is due to movement of ions and bound charges within tissues and due to these reasons, the dielectric properties of biological tissues are highly dependent on frequency and temperature [19]. Recently, the complex refractive indices of freshly excised, non-fixed, human liver tissues have been experimentally determined at five different wavelengths (450–1551 nm) [20]. The liver tissue samples considered in the study are categorized as normal or non-cancerous (N), metastatic (MET), and primary liver tumor (HCC). Based on the above experimental results, it was described that in the above wavelength range, the complex refractive index ($n_s = n_r + in_i$) of above tissue samples can be written as:

$$\textit{Real}: \quad n_r(\lambda) = A_r + \frac{B_r}{\lambda^2} + \frac{C_r}{\lambda^4} \quad (\lambda \text{ in } \mu\text{m}) \tag{3}$$

$$\textit{Imaginary}: \quad n_i(\lambda) = a\lambda^{1-b} \quad (\lambda \text{ in nm}) \tag{4}$$

In the above expressions, the five coefficients $A_r$, $B_r$, $C_r$, $a$, and $b$ take different values for above-mentioned N, MET, and HCC samples as listed in Table 1.

Fig. 2 represents the spectral variation of real part and imaginary part (in inset) of refractive index of three liver tissue samples in wavelength range 450–1551 nm [20].

The present work reports the design considerations to enable fiber optic sensor-based monitoring of liver tissue samples by making use of the experimental data provided by Gainnios et al. [20]. Chalcogenide fiber with intensity interrogation method has been considered for the proposed scheme. The influence of operating wavelength, design parameters (sensing region length and fiber core diameter), light launching conditions, and different fiber glass materials on proposed sensor's performance is studied in order to identify the best possible working conditions leading to a highly accurate and reliable fiber-optic detection of cancerous liver tissues. Possible physical explanations related to simulation results have been provided.

## 2. Design considerations

In this section, different components that are considered in proposed sensor are discussed in view of their optical properties.

### 2.1. Chalcogenide glass

In this model, we are using $As_{40}Se_{60}$ chalcogenide glass material for core and $Ge_{33}As_{12}Se_{55}$ chalcogenide glass material for cladding of the fiber. The refractive index (*RI*) of chalcogenide glasses is dependent on temperature and wavelength, and is represented in terms of following Sellmeier expression:

$$\mathbf{As_{40}Se_{60}}: \quad n_1(\lambda) = \sqrt{\frac{B_1(T)\lambda^2}{\lambda^2 - C_1(T)^2} + \frac{B_2(T)\lambda^2}{\lambda^2 - C_2(T)^2} + \frac{B_3(T)\lambda^2}{\lambda^2 - C_3(T)^2}} \tag{5}$$

$$\mathbf{Ge_{33}As_{12}Se_{55}}: \quad n_2(\lambda) = \sqrt{A(T) + \frac{B_4(T)\lambda^2}{\lambda^2 - C_4} + \frac{B_5(T)\lambda^2}{\lambda^2 - C_5}} \tag{6}$$

where $\lambda$ denotes the wavelength in μm and $T$ is temperature in Kelvin. Also, $A$, $B_1$-$B_5$, and $C_1$-$C_5$ are corresponding Sellmeier coefficients [21,22]. From the above expressions, it is observed that, the condition of total internal reflection (TIR) (Core *RI* ($n_1$) > Cladding *RI* ($n_2$)) is satisfied at any wavelength. Also, the large difference between $n_1$ and $n_2$ for all wavelengths ensures that the fiber remains multimoded with large numbers of modes (angles). The cladding from a small portion (~few mm) around the central region of the fiber is removed and is brought into contact with the analyte medium (discussed below).

### 2.2. Analyte medium: liver tissue sample

The analyzing medium in the present fiber optic sensor is liver tissue sample. By using the data of Gainnios et al. [20] (Eqs. (3) and

**Table 1**
Values of coefficients to calculate the complex refractive indices of N, MET, and HCC samples (Gainnios et al. [20]).

| Tissue | $A_r$ | $B_r$ (μm$^2$) | $C_r$ (μm$^4$) | a | b |
|---|---|---|---|---|---|
| N | 1.35910 | 0.00827 | −0.000576 | 196 | 1.27 |
| MET | 1.34127 | 0.00634 | −0.000324 | 780 | 1.41 |
| HCC | 1.34348 | 0.00998 | −0.000793 | 145 | 1.14 |