Estimating sugar concentration in human blood serum using Surface Plasmon Resonance (SPR) –based optical fiber sensor

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Abstract

This paper presents the Surface Plasmon Resonance (SPR)-based plastic optical fiber sensor for estimating the concentration and refractive index of sugar in human blood serum. (SPR) response curve for different samples of the blood serum was recorded in this work and exhibited a dip in the position of the resonance.it is found that the change in the refractive index of the sensing medium (blood serum) led to the change in the resonance wavelengths in the (SPR) response curve. The resonance wavelength increase as the refractive index increase and this has been proven work at resonance wavelength (613 nm) the refractive index is (1.3628) and this value increased by increasing the resonance wavelength until it reaches the highest value it is (1.387) at resonance wavelength (734 nm).

حساب تركيز السكرفي مصل دم الانسان باستخدام متحسس الالياف البصرية المستندة الى رنين بلازمون السطح

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الخلاصة

يقدم هذا البحث متحسس الالياف البصرية البلاستيكية المستندة الى رنين بلازمون السطح (SPR) لحساب تركيز ومعامل انكسار السكر في مصل دم الانسان. تم تسجيل منحني استجابة لعينات مختلفة من مصل الدم في هذا العمل وعرض التراجع في موقع الرنين. وجد ان التغير في معامل انكسار الوسط المتحسس (مصل الدم) يؤدي الى التغير في الاطوال الموجية الرنينية في منحني استجابة (SPR). يزداد الطول الموجي الرنيني كلما زاد معامل الانكسار وقد أثبت هذا العمل فعند الطول الموجي الرنيني (613 nm) معامل الانكساريكون (1.3628) وتزداد هذه القيمة بزيادة الطول الموجي الرنيني حتى تصل الى اعلى قيمة وهي (1.387) عند الطول الموجي الرنيني (734 nm).

Introduction

Optical fiber-based sensing technology has achieved an escalating attention because the last few decades due to its many benefits over conventional sensing technologies. Compact size, easy fabrication, flexibility and more stable response are the main advantages of the optical fiber sensors. Moreover, they are immune to electromagnetic interference and can easily be installed for structural health monitoring in harsh environments and remote areas. The optical fiber sensors can also be integrated with the laboratory on chip-based devices and the complex sensing networks [1]. Optical fiber sensors field has increased in its research lines and possibilities with the use of nanocoating deposition techniques. Nanostructured thin films and

nanocoatings have been applied to the diverse optical fiber configurations for the fabrication of new sensors [2]. Optical fiber sensors have been developed to measure strain, temperature, pressure, current, voltage, gas, chemical contaminant, rotation, vibration, acceleration, bending, torsion, displacement, and biomolecules [3]. In 1993 the use of optical fiber for surface plasmon resonance was first presented [4]. Surface Plasmon Resonance (SPR) is a high-sensitivity optical sensing method that is utilized for the real-time detection of little variations in the active refractive index of dielectric-metal interfaces. This method is depended on the interaction between free electrons of the metallic layer and light. When the oscillation frequency of the metal's electrons equals the frequency of incident light the resonance takes place, so the intensity of the light reflected off the metal film reduces significantly, and a surface plasmon wave is formed by photon energy transfer into the metal layer [5-9]. In order to excite

The surface plasmon's an optical element like high refractive index prism, optical fiber and diffraction grating are utilized [10].

The experimental work

The experimental setup for measuring the transferred light spectrum consists of the light source (halogen lamp), optical grade plastic optical fiber from Thorlabs and finally, the optical spectrum analyzer (OSA) which connected to a computer. The experimental setup is shown as a photograph in Fig.1 and as a schematic in Fig.2.
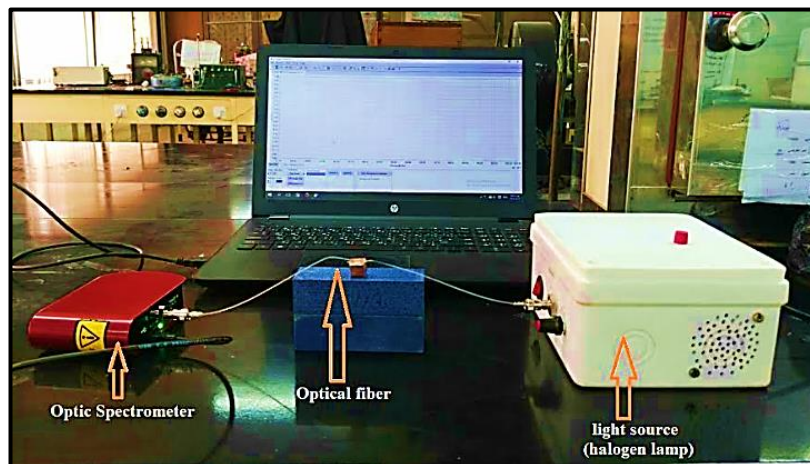

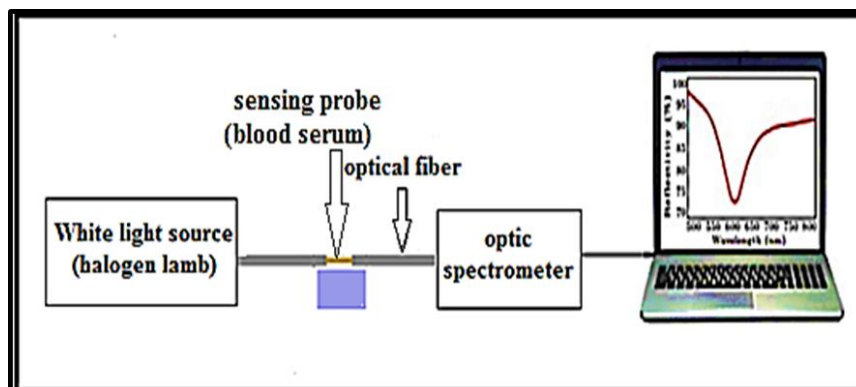*Fig.1: The experimental setup of SPR fiber optic sensor.*

Fig.2: The outline of the experimental setup of SPR based on plastic optical fiber.

A small part of (10 mm) of optical grade plastic optical fiber with a core diameter (980 μm) and numerical aperture (0.51) without a jacket in the middle is embedded in a resin block as shown in Fig.3 and then deposited with about (40 nm) thickness of gold metal by using (ION _COATER) machine from COXEM Company, Korea. The condition used to deposit 40nm thickness of gold was 7 mA current, 300 sec time of deposition. The thickness of the metal layer was determined by Fig.4. The blood serum is placed on gold coated core of an optical grade plastic optical fiber.
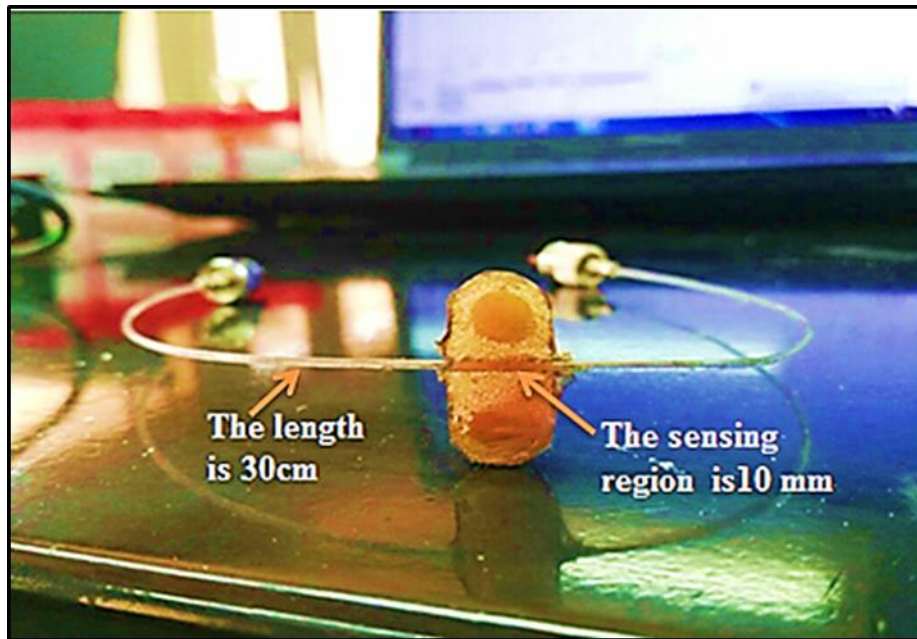

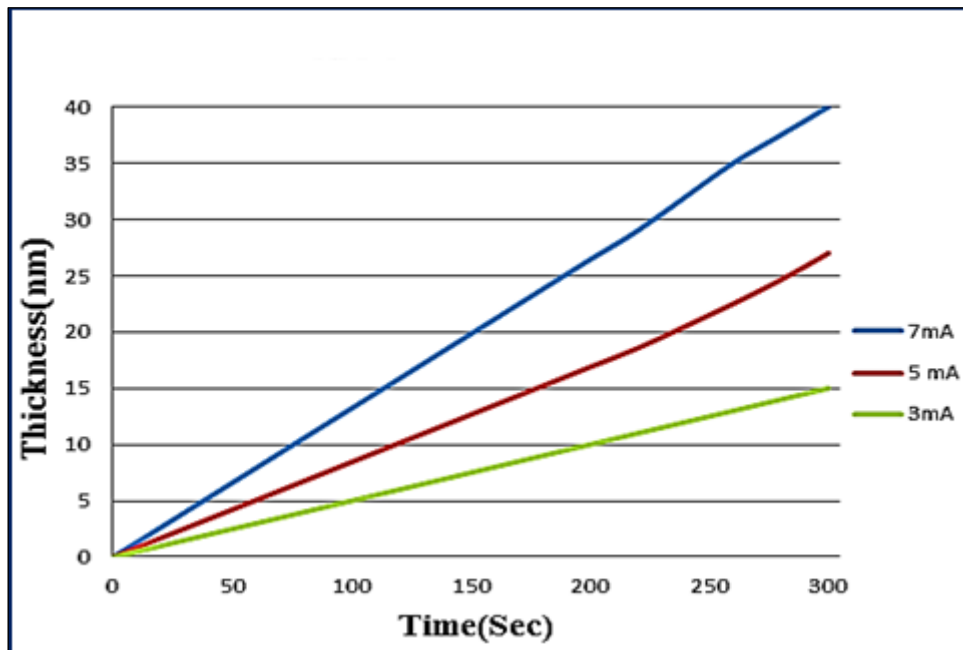*Fig.3: The photograph of the optical grade plastic optical fiber sensor.*

Fig.4: The thickness of the gold layer as a function of time in second.

43

1. Preparations of test solution

The sensitive region of the sensor is covered in various sucrose /water solutions with various concentrations and then different n_s refractive indices to determine the calibration curve as shown in Fig.5. The refractive indices of the solutions were measured by using (Abbe) refractometer. Fig.6 explain the relationship between the refractive index and the solution concentrations.
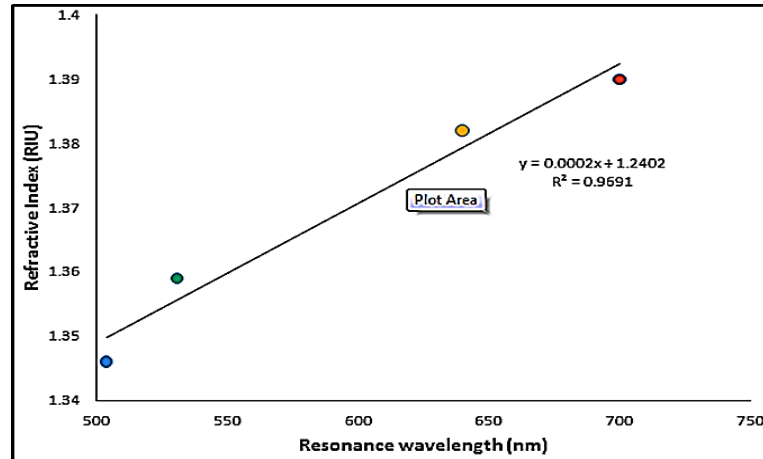

Fig.5: Refractive index as a function of resonance wavelength for the sensor with gold layer. The curve fitting is done on this figure according to equation y=0.0002x+1.2402.
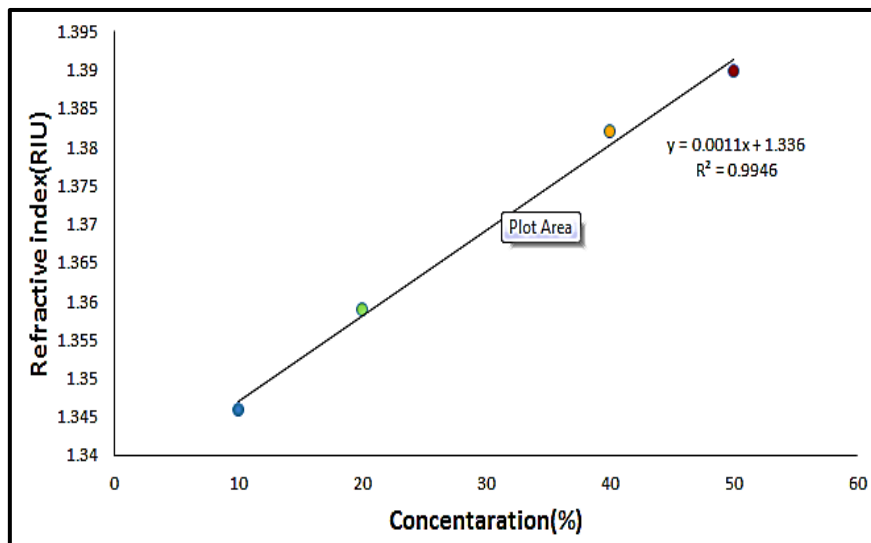

Fig.6: Refractive index of sucrose solutions as a function of the concentration of sucrose. The curve fitting is done on this figure according to equation y=0.0011x+1.336.

2. Preparation of blood serum samples

In this study, 6 serums of blood samples were collected from 6 centrifuged human blood samples. The natural color was yellow for all collected samples. Blood serum samples were used for different people in age ranged from (20 – 40) years. The samples of blood serum are illustrated in Fig.7.
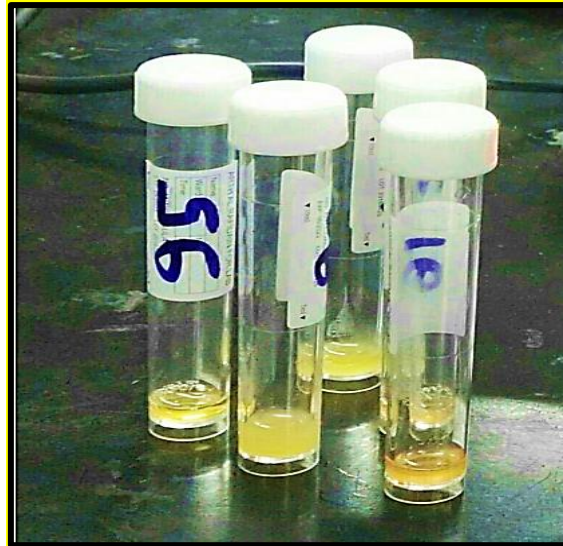
Fig.7: Photograph of the samples of blood serum.

Results and discussion

In this work various values were (1.346, 1.359, 1.382 and 1.39) of refractive index from sucrose /water solutions were found.

The spectra are obtained by recording the transition curves (T) of light through optical fiber. (T) is calculated from the ratio of the intensity (I) measured in the appearance of a sample (sensing medium) and the intensity of the optical signal I_0 that measured in the absence of the sample as in Fig.8. This reference corresponds to the spectrum of the light source "filtered" by all the optical elements of the device Fig.8.
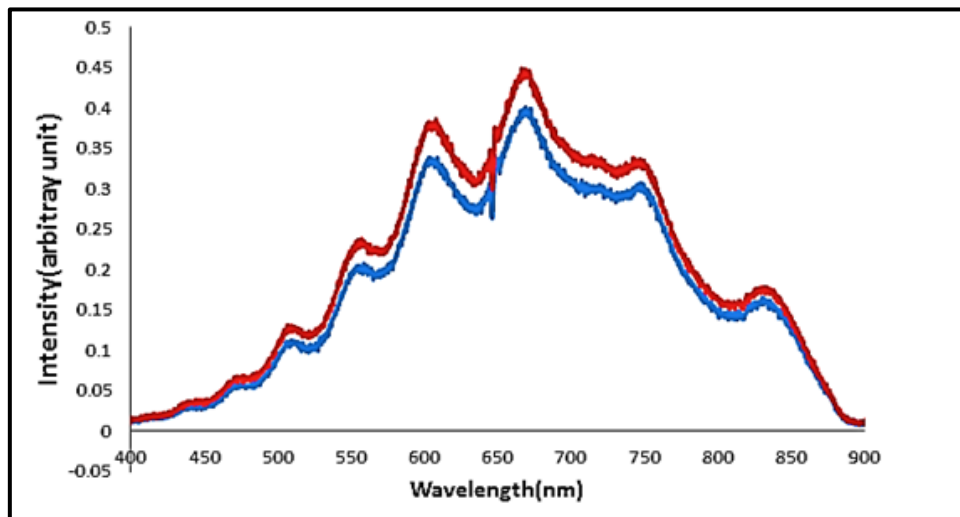

Fig.8: Spectra of the polychromatic light through the sensor measured using the spectrum analyzer; red line for sensitive region of optical fiber in air, blue line for sensitive region dipped in a solution of refractive index n = 1.39.

Fig.9 shows the intensity in the appearance of a sample (sensing medium) should be greater than the intensity in the absence of the sample.

T, calculated as a percentage, is plotted in Fig.9 as a function of the wavelength expressed in nm.
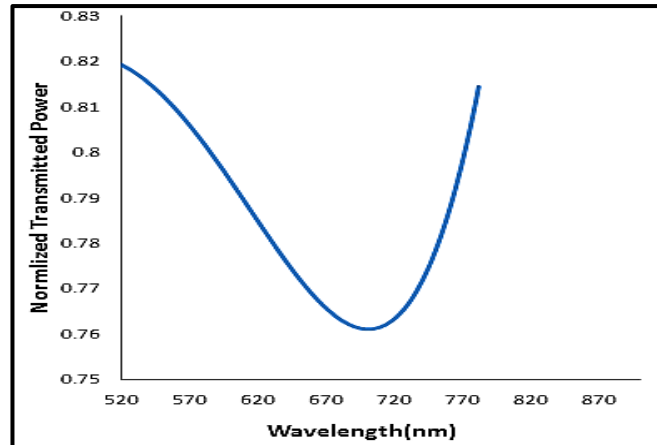
Fig.9: Normalized transmitted power obtained through a sensor fiber, metallized with gold of thickness d=40 nm, L= 10 mm, D= 980 μm) immersed in a solution having an index n = 1.39. This curve is calculated directly from the spectra of Fig.8.

The T-wavelength curve is called SPR curve and at a particular wavelength named resonance wavelength, a sharp dip happens in T because of the energy of incident light transfer to the electrons of the metal and thus reduces the reflected light intensity as shown in Fig.9. The location of this dip depends on the refractive index (n) of the sensing medium if the refractive index of the sensing medium is high the dip will be at high wavelengths, otherwise it will be at low wavelengths. As the refractive index of the sensing medium increase the resonance wavelength increase because the sharp dip shifting to the red wavelength as shown in Fig.5.

To determine the calibration curve, the samples of known refractive indices, which prepared as shown in Fig.6 from different solutions of different concentration of sucrose /water, were used in the fabricated sensor to determined resonance wavelengths for each sample as shown in Fig.5. Then the resonance wavelengths of the unknown sample (blood serum) were determined using the fabricated sensor, and then the refractive indices were determined using the calibration curve in Fig.5. Fig.10 explains the surface Plasmon resonance (SPR) for the fabricated sensor with gold layer at various refractive index of the blood serum (sensing medium). It is obvious that the width and dip position of each surface Plasmon resonance response curve is changed with increased the refractive index to the sensor with each sample having a different refractive index and also the magnitude of shifting the dip position increase as the refractive index increase.
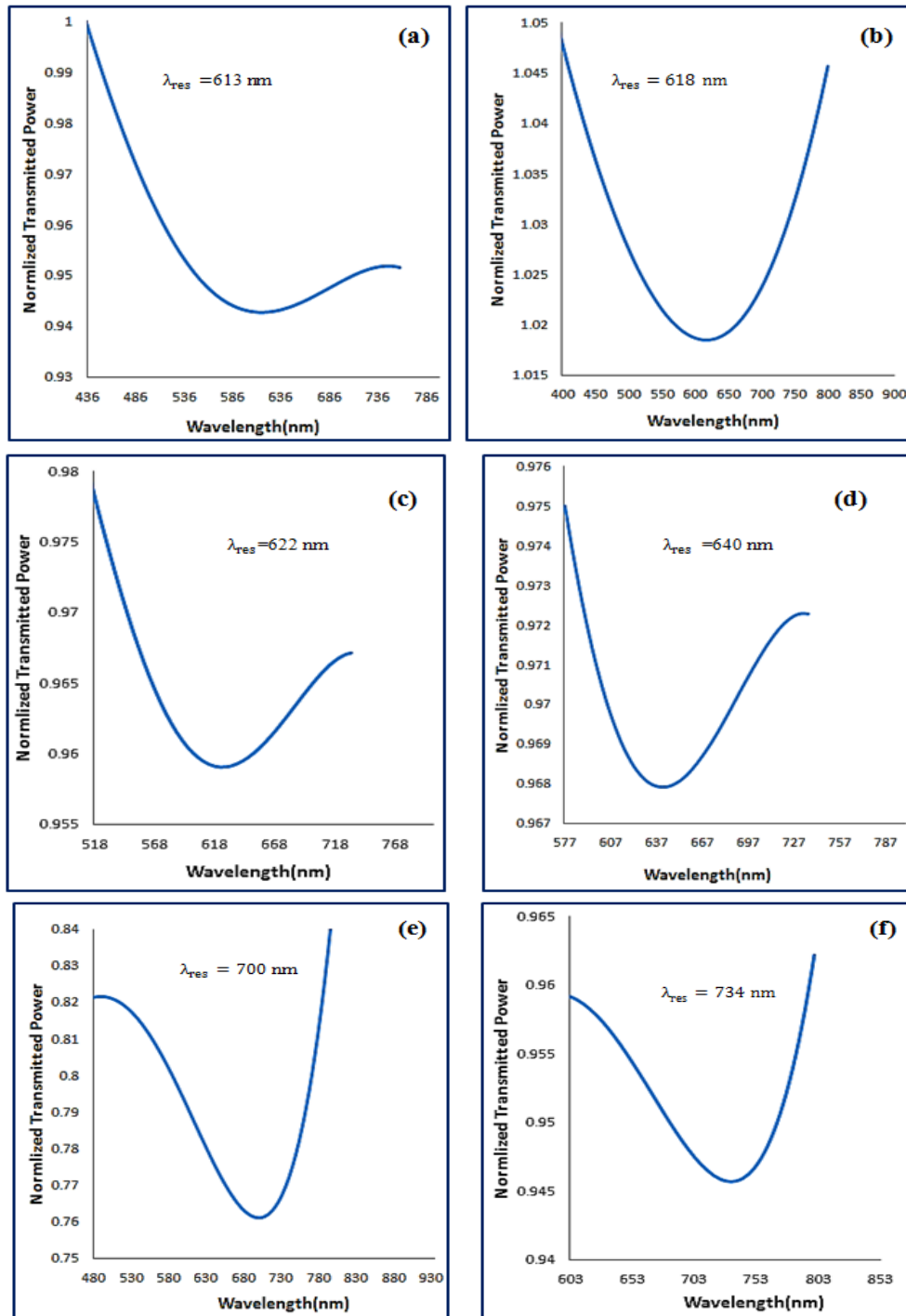
Fig.10: SPR curve of the optical fiber sensor with a gold metal of different samples of blood serum.

The concentration of the samples increases as the refractive index increases and hence the resonance wavelengths increase this happen because the sharp dip shifting to the red wavelength.

Table 1 explains the values of the refractive indices and concentration of sugar for each sample of blood serum at different resonance wavelengths. The resonance wavelengths have been determined from SPR curves for various samples of blood serum in Fig.10 while refractive index and concentration values were calculated from slope equations in Figs.5 and 6

respectively. From this table, it is obvious that the resonance wavelength increase with increasing the sensing medium (blood serum). The benefit of this table is to know the change in the concentration measured by the sensor.

Table 1: Values of the refractive index and concentration for various resonance wavelengths.

Samples	λ_{res} (nm)	Refractive index (RIU)	Concentrations by the sensor	Concentration by laboratory method
a	613	1.3628	24.364	113
b	618	1.3638	25.272	116
c	622	1.3646	26	119
d	640	1.3682	29.273	120
e	700	1.3802	40.182	125
f	734	1.387	46.364	249

Conclusions

This research exhibits the utilizing of optical grade plastic optical fiber as a biomedical sensor based on (SPR) technique for estimating the concentration and refractive index of sugar in human blood serum. The response curve of (SPR) for various samples of blood serum was shown, and a dip in the resonance position was presented in this work. At each sample of blood serum, the resonance wavelength change. The change in the value of the resonance wavelength be happen for each change in the refractive index and then for each concentration of sugar in blood serum. Up to our knowledge this is the first time for producing this high sensitivity sensor for measuring the blood sugar.

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