

Titanium dioxide/zinc telluride/Aluminium antimonide-assisted Label-free Plasmonic sensor for early detection of Glucose in Urine

Manikanta Jetti

Koneru Lakshmaiah Education Foundation

Sandeep Boddu

Koneru Lakshmaiah Education Foundation

Yesudasu Vasimalla

vasimalladasu@gmail.com

Koneru Lakshmaiah Education Foundation

Suman Maloji

Koneru Lakshmaiah Education Foundation

Santosh Kumar

Koneru Lakshmaiah Education Foundation

Research Article

Keywords: Surface plasmon resonance, K108 prism, titanium dioxide, zinc telluride, transfer matrix method, Urine -Glucose detection

Posted Date: July 4th, 2026

DOI: <https://doi.org/10.21203/rs.3.rs-10096383/v1>

License:   This work is licensed under a Creative Commons Attribution 4.0 International License.

[Read Full License](#)

Additional Declarations: No competing interests reported.

Titanium dioxide/zinc telluride/Aluminium antimonide-assisted Label-free Plasmonic sensor for early detection of Glucose in Urine

Manikanta Jetti¹, Sandeep Boddu¹, Yesudasu Vasimalla^{1,*}, Suman Maloji¹ and Santosh Kumar^{1,*}

¹*Centre of Excellence for Nanotechnology, Department of Electronics and Communication Engineering, Koneru Lakshmaiah Education Foundation, Vaddeswaram, Andhra Pradesh–522302, India*

**Correspondence: vasimalladasu@gmail.com and santosh@kluniversity.in*

Abstract: The article presents a high multi-layered surface plasmon resonance (SPR) sensor for the detection of glucose in urine. This designed sensor consists of a K108 (LZOS) prism, a silver (Ag) layer, and dielectric layers of titanium dioxide (TiO_2), zinc telluride (ZnTe), and aluminium antimonide (AlSb) to enhance plasmonic resonance and analyte interaction. The optical response is modeled using the transfer matrix method (TMM) along with angular interrogation at a wavelength of 633 nm. The concentration of glucose in urine affects its refractive index (RI), varying from 1.335 (non-diabetic) to 1.347 (highly diabetic). The numerical simulations based on the TMM revealed a maximum sensitivity of 416.42°/RIU and a quality factor of 142.28 RIU⁻¹, a signal to noise ratio of 1.707, figure of merit of 129.07 and combined sensitivity factor of 91.18, indicating the high resolution and sharp resonance characteristics. The results show that the integration of dielectric and semiconducting materials is significantly enhanced by the plasmonic behavior, giving significant improvement over traditional SPR biosensors. The standard fabrication steps for designing the proposed structure is shown at the end.

Keywords: Surface plasmon resonance, K108 prism, titanium dioxide, zinc telluride, transfer matrix method, Urine -Glucose detection.

1. INTRODUCTION

Detection of glucose in urine is an essential part of patients with metabolic disorders and diabetes mellitus as using clinical evidence test to know if the patient not determining glucose level in urine and what if we have it would reflect in severe complication if not treated [1]. Conventional approaches may rely on invasive blood tests, through the latest advances have developed non-invasive methods based on the use of urine samples. Several new methods have also been developed, such as gold nanoparticles used in microfluidic systems for highly sensitive and selective glucose monitoring [1]. Furthermore, Whispering Gallery Mode (WGM) sensors have been developed to enhance the sensitivity with high-Quality Factor (Q-factor) and linear response to changes in glucose concentration [2]. Other approaches, such as image analysis and molecular imprinting polymers, have also demonstrated potential for non-invasive glucose monitoring which demonstrates the applicability and efficiency of these new technologies [3]. In addition, photonic crystal sensors have been developed for high-sensitive glycosuria detection, thus implying that glucose in urine may be monitored in real-time [4]. Taken together, these innovations represent a trend toward easier to use glucose-monitoring solutions.

Classical methods for glucose detection in urine have improved considerably, with a variety of methods that must be used to guarantee the accuracy and specificity. One of the first methods used was the copper salt reduction, which was not specific for glucose, but for the presence of

reducing sugars in urine [5]. This however was interfered by other substances such as uric acid and creatinine. Enzymatic procedures have subsequently gained greater acceptance, of which the glucose oxidase-peroxidase method is one. It involves a technique used in paper-strip tests, like Clinistix, being specific for glucose can be interfered with by normal urinary constituents, uric acid and ascorbic acid [5]. To overcome these limitations the hexokinase method was adapted as more specific and sensitive method for urine glucose, essentially free of interference of all the other urine factors. Similarly, Clinitest and enzyme dipstick techniques like Diastix and Tes-Tape have been used to monitor glucose concentrations, particularly in diabetics. These approaches have differing sensitivity and specificity, and some can be affected by other compounds such as ketones [6, 7]. These developments underscore the continued quest to develop increasingly accurate and reliable urine glucose measurements.

Surface plasmon resonance (SPR) sensor is a newly developed and extremely sensitive optical technique which detects changes in refractive index [8-10], which happens upon direct interaction of a biological or chemical analyte next to the metal film of the sensor. Despite the fact that several SPR-based biosensors have been developed, the sensitivity of the sensors and their operating spectral region are still subjecting of further improvement [11-13]. The operational principle is based on stimulating surface plasmons synchronized oscillations of free electrons at the boundary between a metal and a dielectric when p-polarized light enters through a prism. Surface Plasmon Resonance (SPR) occurs when the real parts of the dielectric constants of the metal and the adjacent medium differ in sign, enabling the formation of a momentary electromagnetic wave that is tightly confined and decays exponentially into the dielectric layer. This evanescent field sensitivity towards refractive index changes in solution close to the surface, has made the SPR-based sensors efficient for accurate, label-free, and real-time bio-sensing. In a certain angle of incidence, the light couples to the free electrons on the metal surface (plasmons) resulting in the creation of surface plasmon waves (SPW).

The resonance is affected and measured, and the change in resonance is measured, and provides real time detection of molecular interactions. Mostufa *et al.* [14] have designed an SPR-based biosensor, with a gold-thin film deposited on a BK7 glass prism, to detect the haemoglobin in urine. It could achieve acceptable sensitivity, but the single-metal layer limited the resonance sharpness and the detection capability of the system. For performance improvement, multilayer SPR configuration with graphene and silver by Karki *et al.* [15] is reported to inclined enhanced sensitivity by the strong surface plasmon coupling and biocompatible property of graphene. Nevertheless, silver coatings are readily oxidizable, so sensors may become less durable under practical biofluid conditions. Yesudasu *et al.* described a prism type SPR sensor with Ag-Si-BP/MXene structure for the detection of haemoglobin [16]. However, environmentally non-stable nature of the black phosphorus and MXene and their degradation and oxidation in water/humidity environment may pose a substantial impediment for direct implementation of such sensors with durable functionality in the long run. To overcome these drawbacks, in the present study a multilayered prism-based SPR sensor for the selective detection of haemoglobin in urine samples is proposed. Silver (Ag) [17] is used as the metallic layer for the generation of surface plasmons since it possesses an outstanding electrical conductivity and it can sustain strong plasmonic oscillations.

In addition, the dielectric layer is applied to prevent oxidation of the metal surface, as well as increasing the plasmon confinement. In this study, titanium dioxide (TiO_2) is chosen as one dielectric material because of its high refractive index of 2.4 at 633 nm, large band gap of 3.2 eV, and excellent transparency in the visible-near infrared spectra range [18]. Further, TiO_2 has a good thermal stability and chemical inactivity, important for repeatable sensing in

biosensing applications. Another material for this function is zinc telluride (ZnTe) because of its high refractive index of 2.7 at 633nm, and direct bandgap of 2.25eV and reasonable optical and mechanical properties [19]. Moreover, ZnTe is compatible with a wide range of semiconductor systems and possesses efficient wave guiding in multilayer plasmonic structures. By the addition of the active sensing layer aluminum antimonide (AlSb) higher sensitivity and specificity are gained. AlSb is an III–V compound that possesses a relatively narrow direct bandgap (1.6eV) and high refractive index (3.3), enabling strong light–matter interaction and the ability to excite surface plasmons [20]. Its good carrier mobility, high surface activity suggest that it could be one candidate for biosensing, especially for detection of low concentration biomolecules. A heterostructure has been introduced by building metal layers with TiO_2 / ZnTe/ AlSb to overcome drawbacks of conventional SPR sensors by improving stability, sensitivity, and applicability for real-time, non-invasive analyte monitoring.

Motivated by these characteristics, in this paper, we present a new prism based SPR sensor for detection of haemoglobin in urine samples. In this study, we reported the first use of label-free plasmonic haemoglobin biosensor in the urine to detect the presence of haemoglobin, which was based on Titanium dioxide (TiO_2) and Zinc telluride (ZnTe). These materials have excellent characteristics such as wide bandgap, low optical loss, and high RI value, which would help enhance the sensor performance. The highest sensitivity value achieved was 416.42 °/RIU, quality factor (QF) was 142.28 RIU⁻¹ and detection accuracy (DA) was 1.707 and figure of merit (FoM) was 94.75, and combined sensitivity (CSF) was 91.18. Furthermore, a comparative study has been done comparing the proposed sensor with previously reported sensor designs, showing a significant improvement in the sensitivity of the proposed sensor. Moreover, the proposed design contains materials or structures that are suitable for the standard fabrication, enabling the experimental demonstration to be implemented with ease. For a better understanding of the optical response of the sensor, results of electromagnetic field distributions have been studied by Finite-Difference Time-Domain (FDTD) simulation, revealing that strong plasmonic confinement and high light–matter interaction occur at the sensing area.

2. DESIGN AND METHODOLOGY

In this section, we describe the critical design parameters that needed for sensor production and the mathematical modeling frame that used to investigate the reflected intensity from the proposed sensor. Furthermore, it specifies the dominant feeling performance characteristics, enabling to quantify the performance of the sensor in terms of efficiency and reliability.

2.1 Design Considerations

The proposed SPR sensor for glucose detection in urine depends on the utilization of Kretschmann configuration that was demonstrated in Fig 1. This structure can reduce the Kerr effect and improve optical coupling performance. The analysis is carried out at a wavelength of 633 nm, and starts with the K108 (LZOS) prism with refractive index (RI) of 1.5115 as first a layer that allows the efficient coupling of light into the sensor. Similarly, it is possible that, in addition, the sensor sensitivity can be increased by employing a prism made of a lower RI material. The second layer is of silver (Ag) has a complex RI of $0.056206 + 4.27761i$, and is selected for its strong plasmonic character and ability to excite the SPP. After a TiO_2 dielectric layer with the RI of 2.5836 is introduced to enhance light trapping and area enhancement at the metal-rich interface. Furthermore, an ZnTe layer with RI of 2.9887 is introduced to increase the confinement of the refractive area. To enhance the overall performance of a suggested

sensor via depositing a two-dimensional (2D) layer on the top of the dielectric layer. In this work several 2D material are discussed such as black phosphorus, graphene, tungsten disulfide (WS_2), molybdenum disulfide (MoS_2), and $AlSb$. Moreover, urine samples in the sensing medium have different glucose concentrations and corresponding refractive indices of 1.335, 1.336, 1.337, 1.338, 1.341, and 1.347 [21]. These differences are associated to the physiological glucose content which enable the sensor to perceive differences in the resonant conditions useful for glucose detections.

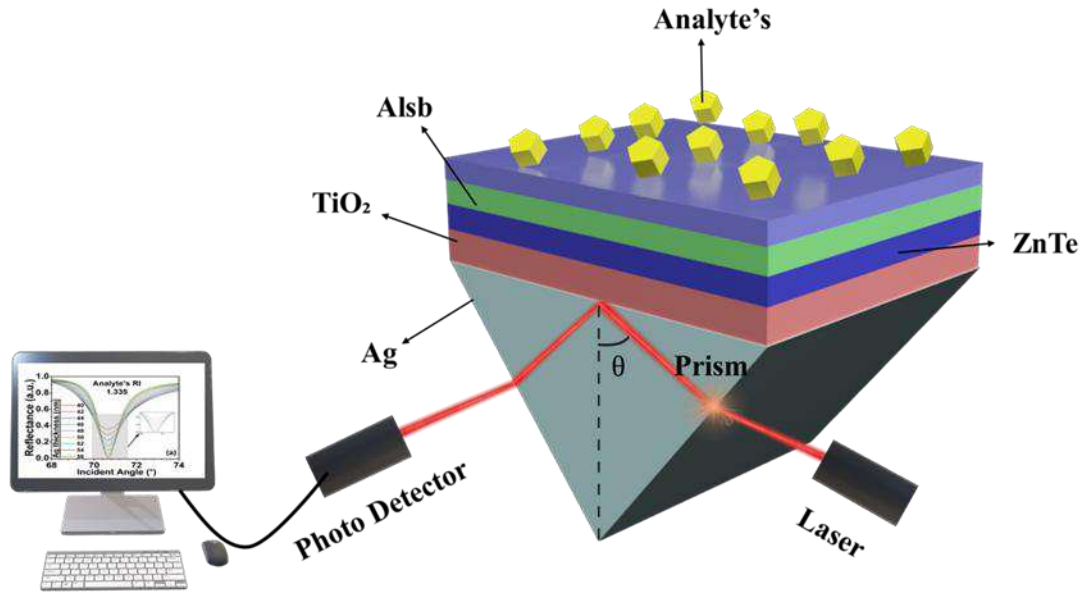


Fig. 1 A schematic diagram of the proposed multilayer SPR sensor.

2.2 Methodology

The Transfer Matrix Method (TMM) is widely used in the modeling of multilayer SPR biosensors involving materials such as K108 (LZOS), Ag, TiO_2 , ZnTe, $AlSb$ and an affinity layer due to its high accuracy and stability. Unlike approximate methods, TMM yields the full sensor optical response, computing the reflectivity across every interface across the multilayered shape. In this apparatus, the SPR is excited while the in-plane portion of the wave vector of the p-polarized incident light (k_x) coincides with that of the floor plasmon wave (k_{SPW}). At the metal-dielectric interface, particularly between silver (Ag) and the adjacent high-refractive index materials including TiO_2 , ZnTe, or $AlSb$. This situation of the resonance is written by Eqs. (1), which govern the coupling behavior and the enhancement of the electromagnetic field in the vicinity of this subject, and enables a highly sensitive detection of the refractive index changes in the glucose-containing urine layer on top of the affinity interface [22].

$$k_x = k_{SPW} \quad (1)$$

The resonance angle (θ_{SPR}) is the angle of incidence at which the minimum reflectance is observed, and represents the optimal coupling of the incident light into the surface plasmon mode. In the above expressions r_p , r_{Ag} , r_{TiO_2} , r_{ZnTe} and r_s are the RI of the K108 (LZOS) prism, silver, TiO_2 , ZnTe, and SM, respectively [23]. These parameters largely determine the optical behavior and performance of the multilayer sensor by affecting the resonance states and total sensitivity.

For the sensor's optical response, we need to use eqns. (2) and (3) for obtaining the affecting factor reflectance (F_p) and its coefficients (f_p) [22]. Meanwhile, Eqs. (4)-(8) present different parameters of the global modelling. In Eq. (6), where the transverse RI of the N^{th} layer is denoted as β_n [22], which describes how light propagates in each layer. Additionally, Eq. (7) is the “characteristic matrix of the integrated sensor structure (Vif)” for p-polarized incident light, which is essential to understand the interaction of light with the multilayer system.

$$F_p = f_p f_p^* = |f_p|^2 \quad (2)$$

$$f_p = \frac{(F_{11} + F_{12}b_N) - (F_{21} + F_{22}b_N)}{(F_{11} + F_{12}b_N) + (F_{21} + F_{22}b_N)} \quad (3)$$

$$b_N \left[\frac{\mu_x}{\varepsilon_x} \right]^{1/2} \cos \theta_x = \sqrt{\frac{\varepsilon_x - (r_p \sin \theta)^2}{\varepsilon_x^2}} \quad (4)$$

$$V_{if} = \left[\prod_{x=2}^{N-1} \begin{pmatrix} \cos \beta_x & -i \sin \beta_x \\ -i n_x \sin \beta_x & \cos \beta_x \end{pmatrix} \right]_{if} = \begin{bmatrix} Z_{11} & Z_{12} \\ Z_{21} & Z_{22} \end{bmatrix} \quad (5)$$

In Eq. (6), β_x is the free phase constant of the absorbing X^{th} layer [24], which has an influence on the wave propagation properties of the incident wave. The input angle (θ_x) and wave impedance parameters (s_x) for this layer are obtained from Equations (7) and (8) [22]. The permeability (μ_x), permittivity (ε_x), and thickness (d_x) of the layer also contribute to the entire optical and electromagnetic characteristics, that is, the wave transmission and reflection inside the multilayer sensor.

$$\beta_x = \frac{2\pi}{\lambda} r_x \cos \theta_x (s_x - s_{x-1}) = \frac{2\pi}{\lambda} d_x \sqrt{\varepsilon_x - (r_p \sin \theta)^2} \quad (6)$$

$$\theta_x = \left(\cos \sqrt{1 - (b_{x-1}/b_x)(\sin \theta)^2} \right)^{-1} \quad (7)$$

$$s_x = \frac{w_x b_k \cos \theta_k}{(2\pi c / \lambda_{633}) \varepsilon_k^2} \quad (8)$$

Eq. (9) gives the reflectivity expression of the proposed five-layered sensor structure in terms of all the relations mentioned above. Here, f_{12} , f_{23} , f_{34} , f_{45} , f_{56} , and f_{67} are the reflected intensities at the interfaces between the different layers [22]. Additionally, d_x and w_x correspond to the thickness and wave vector component of each layer, which have influence on the propagation and resonance condition of the sensor.

$$F_{123456} = |f_{123456}|^2 = \left| \frac{f_{12} + f_{23456} e^{2iw_2d_2}}{1 + f_{12} f_{23456} e^{2iw_2d_2}} \right|^2 \quad (9)$$

$$\text{Where, } f_{23456} = \frac{f_{23} + f_{3456} e^{2iw_3d_3}}{1 + f_{23} f_{3456} e^{2iw_3d_3}}, \quad f_{3456} = \frac{f_{34} + f_{456} e^{2iw_4d_4}}{1 + f_{34} f_{456} e^{2iw_4d_4}}, \quad f_{456} = \frac{f_{45} + f_{567} e^{2iw_5d_5}}{1 + f_{45} f_{567} e^{2iw_5d_5}}, \quad f_{12} = \frac{\varepsilon_2/w_2 - \varepsilon_1/w_1}{\varepsilon_2/w_2 + \varepsilon_1/w_1}, \quad f_{23} = \frac{\varepsilon_3/w_3 - \varepsilon_2/w_2}{\varepsilon_3/w_3 + \varepsilon_2/w_2}, \quad f_{34} = \frac{\varepsilon_4/w_4 - \varepsilon_3/w_3}{\varepsilon_4/w_4 + \varepsilon_3/w_3}, \quad f_{45} = \frac{\varepsilon_5/w_5 - \varepsilon_4/w_4}{\varepsilon_5/w_5 + \varepsilon_4/w_4} \quad \text{and} \quad f_{56} = \frac{\varepsilon_6/w_6 - \varepsilon_5/w_5}{\varepsilon_6/w_6 + \varepsilon_5/w_5}$$

Sensitivity, quality factor (QF), and detection accuracy (DA) are key performance factors for evaluating the performance of a simple SPR sensor. These parameters depend on the FWHM (full-width at half-maximum, the amplitude of the SPR spectrum at 50% reflectivity), change

of the SPR resonance angle ($\nabla\theta_{res}$), and RI change of the analytes (∇n). Mathematically, sensitivity is defined as $\nabla\theta_{res}/\nabla n$, QF as $sensitivity/FWHM$, and DA as $\nabla\theta_{res}/FWHM$. These metrics together define the performance, resolution, and sensing capabilities of the sensor. This optimization of these parameters is very important for improving the sensitivity of SPR sensors for biology, and biomedicine applications with increased sensitivity, and accuracy in analyte detection.

3. RESULTS AND DISCUSSION

In this section we mainly discuss about the simulation results of a proposed sensor by using MATLAB Multiphysics tools. By using an angular interrogation technique at a wavelength of 633 nm. The sensor's performance is studied using the SPR curve, which shows the relationship between reflectance and incidence angle. Mainly, the iteration method can be used to demonstrate the impact of Ag thickness and the addition of the proposed structure. After, the glucose detection technique is then carried out at multiple concentrations using the optimized structure. The sensor's field distribution along with a specific fabricating technique is also analyzed. Finally, the comparison of the sensor's performance with previous studies has been provided.

3.1 Optimal Metallization of layer thickness

The performance of the sensor was studied for different Ag layers on top to determine the ideal metallization layer. In Table I the Ag layer thickness was varied from 40 to 56 nm in 2 nm increment by the iterative technique and the thicknesses of TiO_2 and ZnTe layers were kept as 1 nm, respectively. In Fig. 2(a) and 2(b), which shows the reflectance verses angle of incidence for refractive indices of 1.335 and 1.347 respectively, the resonance angle shifted to higher angles with increasing analyte, such a result in agreement to sensibility to changes in RI. In addition, the larger the Ag thickness, the smaller the minimum reflectance (R_{min}) and the narrower the FWHM, indicating sharper and stronger resonance dips. The maximum sensitivity of 144.14 °/RIU at Ag thickness of 56 nm that is then chosen for the remaining simulation work.

In Table II after the optimization of the metal layer, the effect of the TiO_2 dielectric thickness, which varying from 1 to 8 nm was investigated at a constant Ag thickness of 56 nm and ZnTe thickness of 1 nm. In Fig. 2(c) and 2(d), the resonance angle and sensitivity increased significantly with the TiO_2 thickness increasing from 144.14 to 416.42°/RIU due to stronger plasmonic coupling with improved light confinement and higher refractive index difference. Although R_{min} and FWHM slightly increased, it was not harmful to the performance of the sensor, which achieved the maximum QF of 135.80 RIU^{-1} as well as the FoM of 129.07 at 8 nm thickness of TiO_2 , thus indicating the optimized performance achieved. Although thicker TiO_2 would lose transducer sensitivity in some high sensitivity region, while the thickness of 6 to 8 nm showed the best potential for practical biosensing applications.

FoM	1.335	64.14	66.44	70.12	74.55	80.85	88.21	96.32	104.65	117.41
	1.347									
CSF	1.335	63.49	65.87	69.60	74.07	80.38	87.75	95.87	104.19	116.92
	1.347									

TABLE II

The measured parameters using thicknesses of TiO_2 metal layer for Fig. 1(c) and (d).

	$\theta_{res}(^\circ)$	1nm	2nm	3nm	4nm	5nm	6nm	7nm	8nm
θ_{SPR}	1.335	70.6777	71.5746	72.6637	73.8809	75.2262	76.8919	78.8779	81.5045
	1.347	72.4074	73.4324	74.6496	75.995	77.6607	79.6466	82.3373	86.5015
R_{min}	1.335	0.00158	0.00383	0.00229	0.00396	0.00561	0.00909	0.0190	0.0494
	1.347	0.00382	0.00286	0.00569	0.0066	0.0123	0.0251	0.0671	0.3340
R_{max}	1.335	0.9961	0.9959	0.9955	0.9950	0.9941	0.9924	0.9884	0.9750
	1.347	0.9958	0.9955	0.9949	0.9939	0.9920	0.9870	0.9749	0.9749
FWHM	1.335	1.1259	1.2609	1.4197	1.6129	1.849	2.1482	2.5395	3.0665
	1.347	1.2432	1.3995	1.5887	1.8219	2.1171	2.5046	3.0362	2.9266
$\nabla\theta_{res}(^\circ)$	1.335	1.7297	1.8578	1.9859	2.1141	2.4345	2.7547	3.4594	4.9970
	1.347								
Sensitivity ($^\circ/RIU$)	1.335	144.14	154.82	165.49	176.18	202.88	229.56	288.28	416.42
	1.347								
QF (RIU^{-1})	1.335	128.02	122.78	116.57	109.23	109.72	106.86	113.52	135.80
	1.347								
SNR	1.335	1.5362	1.4733	1.3987	1.3106	1.3166	1.2822	1.3621	1.6294
	1.347								
FoM	1.335	127.81	122.30	116.29	108.79	109.10	105.88	111.35	129.07
	1.347								
CSF	1.335	127.27	121.75	115.73	108.13	108.29	104.47	108.46	125.70
	1.347								

3.2 Optimization of Structure

In Table III the structural optimization of the SPR sensor, R_{min} was considered along with sensitivity and resonance angle shift, and finally R_{min} was chosen as the most effective parameter to investigate the performance for refractive index change of 0.012 for the five structures of S1-S5. From Fig. 3 the highest sensitivity was obtained for structure S5 combination K108/Ag/ TiO_2 /ZnTe/AlSb is $416.42^\circ/RIU$, corresponding at maximum resonance angle shift of 4.997° , but also higher R_{min} values 0.049485 for SM's RI of 1.335 and 0.33404 for SM's RI of 1.347 reveal that higher reflectance exists at resonance. Joint structure S3 had a good sensitivity of $224.23^\circ/RIU$ with a 2.6907° shift, but R_{min} was increased. The S2 and S4 structures showed more balanced performance with sensitivities of 136.47 and $128.13^\circ/RIU$ and lower R_{min} , which indicates efficient plasmon excitation with less energy loss. S1 showed the minimum sensitivity of $122.78^\circ/RIU$ while holding the smallest value of R_{min} . These results demonstrate that although S5 offers the best sensing performance, designs such as S2 and S4 enable a balance between the sensitivity and reflectance loss, which is important for

practical and robust sensing applications.

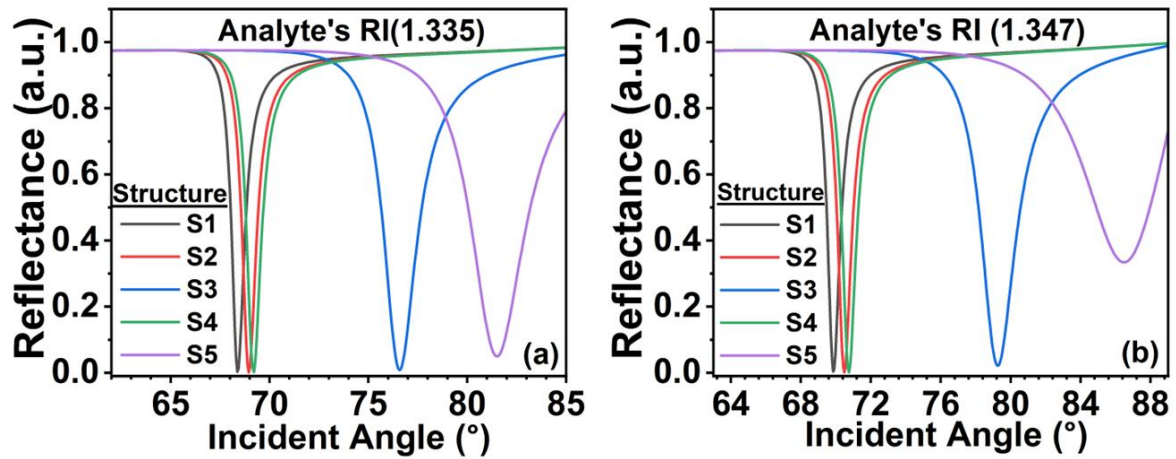


Fig. 3(a) Reflectance versus angle at SM's RI of 1.335 for different structures and (b) Reflectance versus angle at SM's RI of 1.347 for different structures.

TABLE III

Obtained Parameters of R_{min} , θ_{res} , R_{max} , FWHM and $\nabla\theta_{res}$ for Fig. 2

Structure	R_{min}		θ_{res} (°)	θ_{res} (°)	$\nabla\theta_{res}$ (°)	Sensitivity (°/RIU)
	1.335	1.347	1.335	1.347		
S1	0.0032127	0.0038815	68.3714	69.8448	1.4734	122.78
S2	0.0011065	0.0013124	68.9479	70.4855	1.6376	136.46
S3	0.0078247	0.0211	76.5716	79.2623	2.6907	224.22
S4	0.0019781	0.0012915	69.2042	70.7417	1.5375	128.12
S5	0.049485	0.33404	81.5045	86.5015	4.997	416.42

3.3 Performance analysis for 2D materials

From Fig. 4 two-dimensional (2D) materials are good supplements for boosting in the SPR-based biosensors due to their unique optical, electrical, and surface properties, which is conducive to the non-invasive detection of urine glucose as well. In the present work performance of 4 different 2D materials, i.e. AlSb, BP, Graphene, WSe_2 have been examined over an RI range of 1.335-1.347 (which ultimately varies the glucose concentration in urine). The simulation results from the origin plots indicate that AlSb thickness of 0.53 nm, one atomic layer showing the highest sensitivity up to 416.42°/RIU, provides sharp resonance dips, and has excellent plasmonic field enhancement. BP possessed a moderate-high sensitivity with heavily dispersive resonant behaviours for the tunable bandgap and anisotropic optical performance, so that it was a promising candidate for field confinement improvement. Graphene, which had a little lower sensitivity, was not only sensitive but also has the stable and linear shift of resonance across the whole RI region by possessing excellent electric property and molecular adsorption. WSe_2 , another TMD motif, presented a good sensitivity at higher RI of 1.341–1.347 due to its large refractive index contrast and excitonic effects. Based on the above-mentioned RI vs. resonance angle graph of origin, it is clear that all 4 h-BNs

enhanced the performance of the SPR sensor response AlsB is the most sensitive/acute-edge 2D-material, then followed by GeS and ZnS from our sensors, respectively, which qualify it as a novel & promising 2D-material for real-time and mandatory glucose testing at an earlier level in urine.

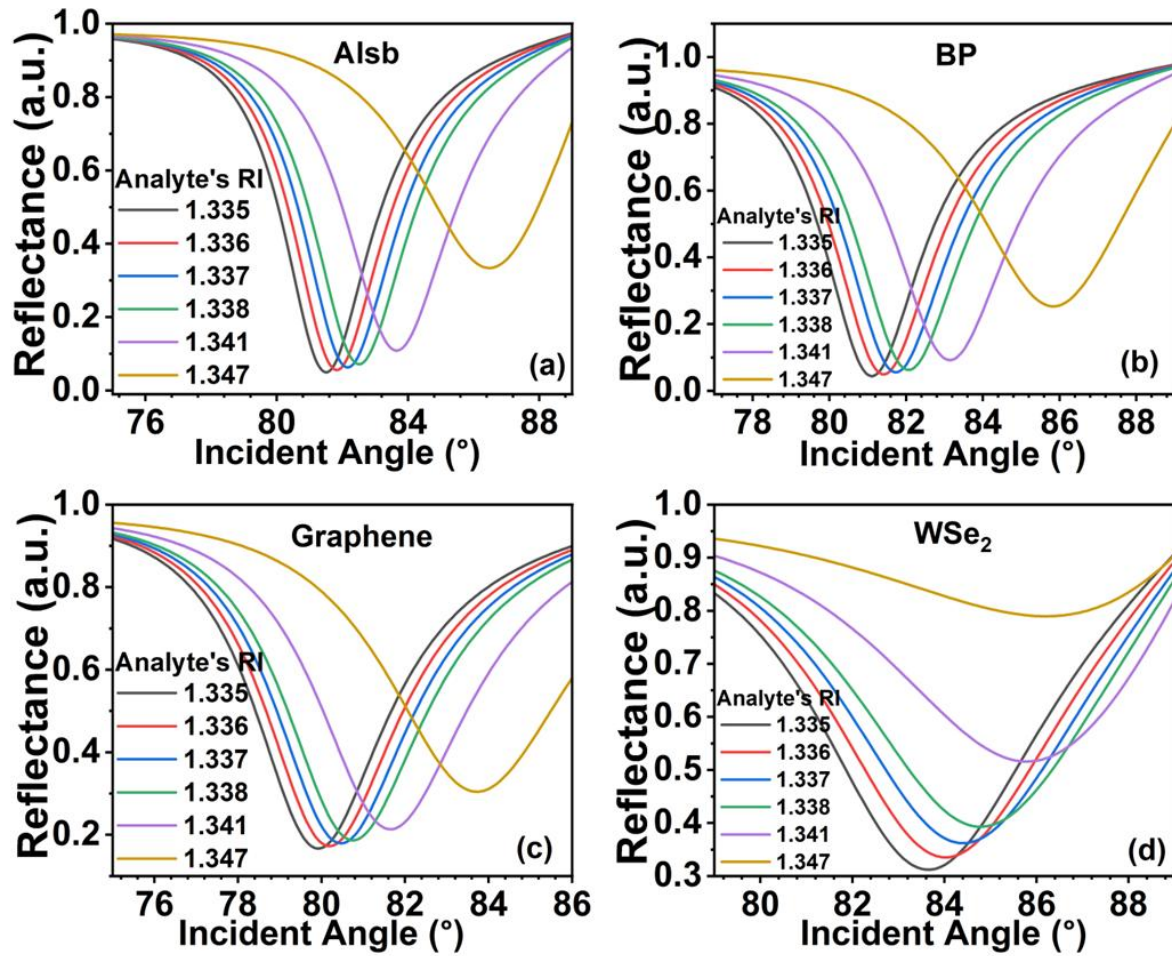


Fig. 4 2D materials of analytes of RI from (1.335-1.347): (a) Reflectance versus angle at SMs RI for AlsB structure and (b) Reflectance versus angle at SMs RI for BP structure (c) Reflectance versus angle at SMs RI for Graphene structure and (d) Reflectance versus angle at SMs RI for WeS₂ structure.

Table IV Obtained basis elements for AlsB-based structure

SM's RI	Sensitivity (°/RIU)	QF (RIU ⁻¹)	SNR	FoM	CSF
1.335	-	-	-	-	-
1.336	320.3	102.6602564	0.102660256	96.94187481	94.3753684
1.337	320.3	100.8628291	0.201725658	94.50403288	91.98246215
1.338	341.67	105.6908054	0.317072416	98.11689661	95.47462647
1.341	363.03	106.6051957	0.639631174	94.98736145	92.31157103
1.347	416.42	142.2868402	1.707442083	94.75734413	91.18594444

From Table IV the sensitivity of the sensor increased with an increase in the refractive index of the sensing medium from 1.336 to 1.347 and reached as high as 416.42°/RIU at RI of 1.347. The QF also increased dramatically from 102.66 to 142.29 RIU⁻¹, and the SNR increased from 0.10 to 1.71, demonstrating that the sensitivity and accuracy of detection were improved. The maximum FoM and CSF were found to be 98.12 and 95.47 at RI of 1.338, demonstrating an impressive robustness of the sensor over different analyte concentrations.

Table V Obtained basis elements for BP-based structure

SM's RI	Sensitivity (°/RIU)	QF (RIU^{-1})	SNR	FoM	CSF
1.335	-	-	-	-	-
1.336	320.3	105.9158097	0.10591581	100.668528	98.0206333
1.337	320.35	104.1246831	0.208249366	98.3386827	95.7355656
1.338	320.33	102.3134988	0.306940496	95.9269878	93.3589190
1.341	341.68	103.4776903	0.620866142	93.9364263	91.3391363
1.347	395.07	118.499855	1.42199826	88.5193917	85.5450453

With the increase of the sensing medium's RI from 1.336 to 1.347, in Table V the sensitivity of the SPR sensor enhanced and increased to 395.07°/RIU at RI of 1.347. Enhancement in QF and SNR also reached 118.50 RIU^{-1} and 1.42, respectively. Though, FoM and CSF slightly dropped for higher RIs, their maximum values of 100.67 and 98.02 were located at RI of 1.336 establishing a good overall performance.

Table VI Obtained basis elements for Graphene-based structure

SM's RI	Sensitivity (°/RIU)	QF (RIU^{-1})	SNR	FoM	CSF
1.335	-	-	-	-	-
1.336	320.3	100.4390091	0.100439009	83.08917466	80.75898965
1.337	288.3	89.73481076	0.179469622	73.65074328	71.38942605
1.338	277.6	85.64200654	0.25692602	69.66292096	67.45335719
1.341	288.28	86.76438131	0.520586288	68.25146527	66.00426779
1.347	320.32	97.98613235	1.175833588	68.16209325	65.61445381

Furthermore, a different multilayer SPR sensor structure was studied in the RI range from 1.336 to 1.347, in order to examine the multilayer SPR sensor sensitivity to different analytes. From Table VI the sensor showed the highest sensitivity of 320.3°/RIU at RI of 1.336 and 1.347, with QFs of 100.44 RIU^{-1} , 97.99 RIU^{-1} , respectively. Although in the intermediate values of RI the sensitivity slightly lowered the global trend shows that it can provide a good sensitivity to RI variations. The SNR increased steadily up to 1.1758 at RI of 1.347 and it would indicate that the resolution would be better at the higher analyte concentration than at a lower concentration. Figure of Merit (FoM) and Contrast Sensitivity Factor (CSF) exhibited a similar trend in variation, with FoM of 83.09 and CSF of 80.76 at RI of 1.336. These results show that this structure still exhibits efficient performance over a clinically useful range of RI, but is somewhat less sensitive than the 2D material-integrated design. It continues to show potential as an accurate urine glucose analyser.

Table VII Obtained basis elements for WeS_2 -based structure

SM's RI	Sensitivity (°/RIU)	QF (RIU^{-1})	SNR	FoM	CSF
1.335	-	-	-	-	-
1.336	384.4	113.1886576	0.113188658	75.24781956	72.33887106
1.337	384.4	120.1738206	0.240347641	76.67810798	73.57762341
1.338	384.4	133.3009675	0.399902903	80.92701737	77.48785241

In order to further improve the sensing performance of the device, another combination of 2D materials was examined by simulating the response of the sensor in response to varying RI of the analyte. From Table VII the sensor exhibited reproducible and remarkable performance with a constant sensitivity of 384.4°/RIU in the RI range from 1.336 to 1.338. The QF improved from 113.19 RIU^{-1} at the reference RI of 1.336 to 133.30 RIU^{-1} at the RI of 1.338, which means the resonance curve is even more sharp and clear with the increased of RI. Besides, other relevant metrics as the signal-to-noise ratio (SNR), figure of merit (FoM), and contrast sensitivity factor (CSF) also increased in a relevant manner, evidencing the success of the

sensor layout.

3.4 Comparative study on Performance analysis

A performance comparison between the previously reported SPR sensors for urine glucose sensing is presented to confirm the viability of the developed sensor architecture. From Table VIII the previous systems utilised configurations such as FK514/Au/graphene (2020) and BK7/Ag/MXene/ZnO/graphene (2022) which realised sensitivities of $275^{\circ}/\text{RIU}$ and $133.5^{\circ}/\text{RIU}$, respectively, with incomplete details provided on quality factors. Although the BP based structures (e.g., BK7/Au/ZnO/Ag/BP in [2022] and BK7/Au/Si/BP in [2023]) displayed enhanced sensitivity achieved to be $289^{\circ}/\text{RIU}$ and $273.4^{\circ}/\text{RIU}$, sensitivity couplings were outperformed relative to the investigation related to the limit miniaturization as mediated by the quality factor. Newer structures like BK7/Ag/ZnSe and BK7/Au/Ag/AlN/BP showed higher sensitivities of $366^{\circ}/\text{RIU}$ and $394.44^{\circ}/\text{RIU}$ and reported a quality factor of 112.6 RIU^{-1} , respectively. By comparison, the novel sensor (K108/Ag/ TiO_2 /ZnTe/AlSb) shows the highest sensitivity ($\approx 416.42^{\circ}/\text{RIU}$) and a higher quality factor ($\approx 135.80 \text{ RIU}^{-1}$) when compared to all previous designs. The better performance is due to the synergistic effect of AlSb acting as a 2D material, which can effectively enhance plasmonic confinement and field interaction at the sensing interface. These results verify that the designed sensor possesses a great prospect of serving as an effective and credible strategy for early-stage, non-invasive glucose identification in urine.

TABLE VIII Comparative study of performance analysis

Author	Structure	Sensitivity ($^{\circ}/\text{RIU}$)	Quality factor (RIU^{-1})	Reference
Panda <i>et al.</i> (2020)	FK514/Au/Graphene/ Urine sample	275	76	[25]
Karki <i>et al.</i> (2022)	BK7/Ag/MXene/ ZnO/Graphene/ urine sample	133.5	-	[26]
Yadav <i>et al.</i> (2022)	BK7/Au/ZnO/ Ag/BP/Urine sample	289	-	[27]
Yadav <i>et al.</i> (2023)	BK7/Mxene/ ZnO/graphene/ Urine sample	184	-	[28]
Kumar <i>et al.</i> (2023)	BK7/Ag/ZnSe/ Urine sample	366	66	[29]
Karki <i>et al.</i> (2023)	BK7/Au/Si/Bp/ Urine sample	273.4	-	[30]
Karki <i>et al.</i> (2023)	BK7/Ag/Si/ WS2/Urine Sample	200	-	[30]
El-assar <i>et al.</i> (2024)	BK7/Cu/ graphene/ Urine sample	180	-	[31]
Rafi <i>et al.</i> (2024)	BK7/Au/Ag/ AlN/BP/Urine sample	394.44	112.6	[32]
-	K108/Ag/ TiO_2 / ZnTe/Alsb/Analyte	416.42	142.28	This Work

3.5 Stepwise Fabrication Process

The fabrication process to create the proposed SPR sensor for urine-glucose detection in pre-diabetic and early-diabetic states, as confirmed experimentally, is shown in Fig. 5 a K108-type high-refractive-index prism serves as a superior substrate for light coupling. The surface of the prism is cleaned with methanol, acetone, and deionized water to remove the surface impurities and achieve better layer bonding. A thickness of 56 nm-thick film of silver (Ag) is coated on the prism via electron-beam deposition in a high-vacuum system to make the plasmonic layer. An 8 nm thick TiO_2 dielectric layer is deposited by atomic layer deposition (ALD) to improve the electric field confinement and structural integrity. This is followed by 1 nm of ZnTe deposited by physical vapor deposition (PVD) to increase the interface sensitivity and raise the refractive index contrast. Thereafter, as a 2D material, a 0.53 nm monolayer of AlSb is transferred or grown on the ZnTe layer by a chemical vapor deposition (CVD) method. This layer is critical in enhancing plasmonic interactions at the sensing interface. The structure and morphological characteristics of the deposited layers can be verified by using XPS, AFM, SEM and Raman spectroscopy. Finally, synthetic or collected urine samples with glucose concentrations representing pre-diabetic and early-diabetic states (RI values varying between 1.335 and 1.347) are flown over the surface of the AlSb sensor. The resonance angle shifts, with angular interrogation at 633 nm, are monitored for quantitative, noninvasive sensing of glucose changes in model diabetic solutions.

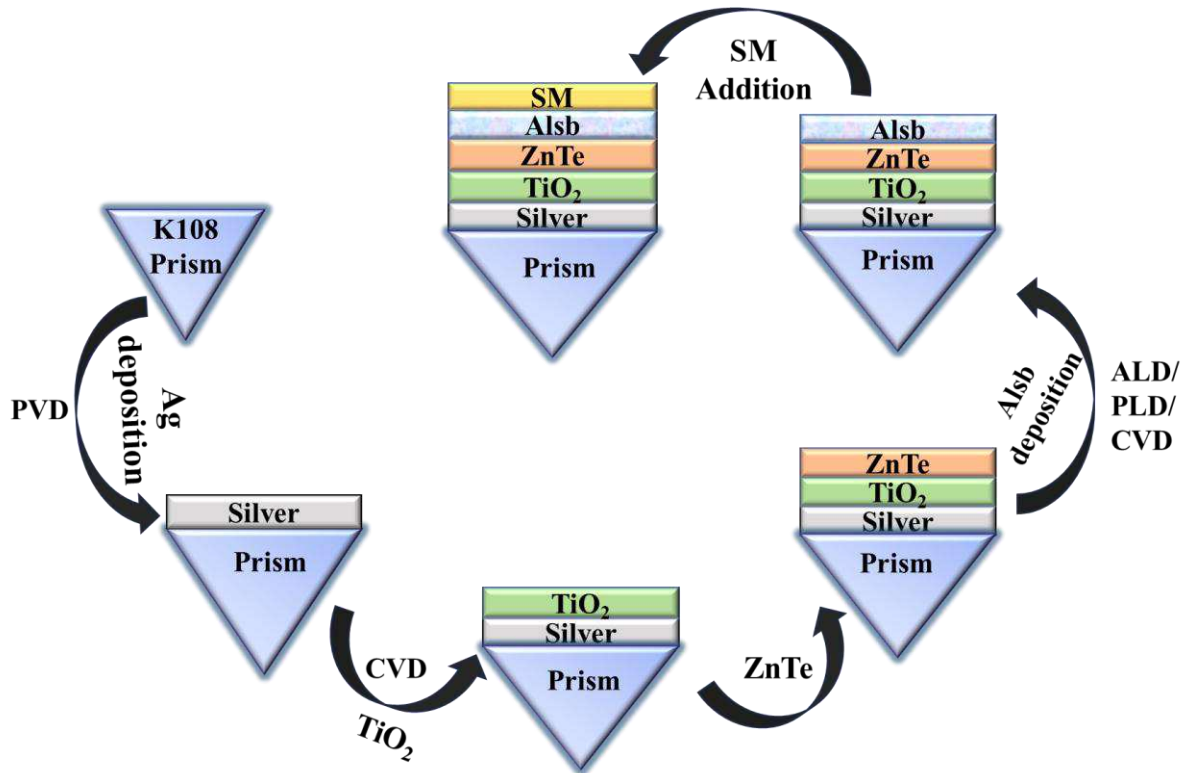


Fig. 5 Steps to fabricate the proposed sensor for analyzing the urine glucose detection.

4. CONCLUSION

A design and performance analysis of a highly performed multilayer SPR sensor for detection of urine glucose to aid in early-stage diabetes detection is presented in this paper. The proposed consists of Ag- TiO_2 -ZnTe-based structure that placed in between prism and sensing medium. In first step, we optimized the considered layers' thicknesses, such as Ag of 56 nm, TiO_2 of 8 nm and ZnTe of 1nm, using the iteration method. Subsequently, in the second step, the impact of the proposed structure is shown, confirming that 3.39, 3.05, 1.86, 3.25 times higher compared to the S1, S2, S3 and S4, respectively, which are divided with the defined layers.

Moreover, in the third part, the performance analysis of the proposed structure is executed and found that the maximum obtained sensing parameters are a sensitivity of 416.42°/RIU, a QF of 142.28 RIU⁻¹, a SNR of 1.707, FoM of 129.07 and CSF of 91.18. This result clearly demonstrating the noteworthy enhancement compare to the existing works. To support practical implementation, fabrication process is also discussed at end. Therefore, this function highlights the possibility of a proposed sensor for real-time, label-free, and cost-effective glucose checks in biomedical diagnostics.

Declarations

Proclamations: Contributions from the authors: M.J., S.B., and Y.V. helped with the manuscript's design, methodology, formal analysis, inquiry, and validation. Conceptualisation, resource provision, supervision, manuscript review, and editing were all done by S.M. and S.K.

Funding: This work was supported by the Koneru Lakshmaiah Education Foundation, India.

Data availability: The datasets utilized and/or analysed in this study can be obtained from the corresponding author upon reasonable request.

Conflict of interest: The authors confirm that there are no conflicts of interest associated with this work.

Ethical approval: This article does not contain any studies involving human participants performed by any of the authors.

Consent to Publish declaration: Not applicable.

Consent to Participate declaration: Not applicable.

References

- [1] K. F. S. Putri, H. Sulistyarti, and A. Sabarudin, "Urine Glucose Detection Via Gold Nanoparticle Formation Using 3D-Connector Microfluidic Paper Based Analytical Devices," *al Kimiya: Jurnal Ilmu Kimia dan Terapan*, vol. 11, no. 1, pp. 51-60, 2024.
- [2] N. A. Merabet and L. Cherbi, "Optimization of a high-performance wgm sensor design for glucose detection in urine," in *2023 International Conference on Numerical Simulation of Optoelectronic Devices (NUSOD)*, 2023: IEEE, pp. 33-34.
- [3] A. Yudhana, L. Y. Sabila, A. C. Subrata, H. H. Pratama, and M. S. Akbar, "Non-Invasive Approach for Glucose Detection in Urine Quality Using Its Image Analysis," in *Biosignal Processing: IntechOpen*, 2022.
- [4] P. Sharma and P. Sharan, "An analysis and design of photonic crystal-based biochip for detection of glycosuria," *IEEE Sensors Journal*, vol. 15, no. 10, pp. 5569-5575, 2015.
- [5] M. Nooshnab, S. M. Naghib, R. Rahmanifard, and E. Khakpour, "A review of minimally and non-invasive glucose monitoring techniques, devices and sensors," *Current Analytical Chemistry*, vol. 20, no. 4, pp. 217-241, 2024.
- [6] B. J. Stubbs *et al.*, "On the metabolism of exogenous ketones in humans," *Frontiers in physiology*, vol. 8, p. 300397, 2017.
- [7] M. C. Nahata, "Cephalosporins," in *Recent Developments in Therapeutic Drug Monitoring and Clinical Toxicology*: CRC Press, 2023, pp. 117-122.

- [8] S. Vankalayapati *et al.*, "Design of a Dual-Metal Layer SPR Biosensor for Enhanced Melamine Detection Using Black Phosphorus Nanomaterials," *Plasmonics*, pp. 1-12, 2024.
- [9] V. V. Kotha *et al.*, "Ultra-Sensitive Prism-Based Surface Plasmon Resonance Biosensor Utilizing Lead Molybdate and BlueP/TDMC Nanocomposites for Early Malaria Detection," *Plasmonics*, pp. 1-16, 2024.
- [10] J. Homola and M. Piliarik, *Surface plasmon resonance (SPR) sensors*. Springer, 2006.
- [11] H. C. Gomes *et al.*, "Laser-Induced graphene-based Fabry-Pérot cavity label-free immunosensors for the quantification of cortisol," *Sensors and Actuators Reports*, vol. 7, p. 100186, 2024.
- [12] R. Singh, W. Zhang, X. Liu, B. Zhang, and S. Kumar, "WaveFlex Biosensor: MXene-Immobilized W-shaped Fiber-Based LSPR sensor for highly selective tyramine detection," *Optics & Laser Technology*, vol. 171, p. 110357, 2024.
- [13] R. Jha, P. Mishra, and S. Kumar, "Advancements in optical fiber-based wearable sensors for smart health monitoring," *Biosensors and Bioelectronics*, p. 116232, 2024.
- [14] S. Mostufa, A. K. Paul, and K. Chakrabarti, "Detection of hemoglobin in blood and urine glucose level samples using a graphene-coated SPR based biosensor," *OSA Continuum*, vol. 4, no. 8, pp. 2164-2176, 2021.
- [15] B. Karki, A. Jha, A. Pal, and V. Srivastava, "A novel design for glucose detection in urine samples using SPR sensor," 2022.
- [16] Y. Vasimalla and H. S. Pradhan, "Modeling of a novel SCHOTT B270 prism based SPR sensor using Ag-Si-BP/MXene structure for detection of specific biological samples," *Optical and Quantum Electronics*, vol. 54, no. 10, p. 612, 2022.
- [17] Y. Li, Q. Liao, W. Hou, and L. Qin, "Silver-based surface plasmon sensors: fabrication and applications," *International journal of molecular sciences*, vol. 24, no. 4, p. 4142, 2023.
- [18] S. Srivastava, S. Singh, A. C. Mishra, P. Lohia, and D. Dwivedi, "Numerical study of titanium dioxide and MXene nanomaterial-based surface plasmon resonance biosensor for virus SARS-CoV-2 detection," *Plasmonics*, vol. 18, no. 4, pp. 1477-1488, 2023.
- [19] S. Singh, S. Singh, P. K. Singh, R. K. Yadav, P. Lohia, and D. Dwivedi, "Theoretical study of malaria detection in blood samples using bimetal layer and zinc telluride nanomaterial-based surface plasmon resonance biosensor," *Plasmonics*, vol. 18, no. 6, pp. 2125-2136, 2023.
- [20] A. Banerjee, R. Dahiya, J. Thangaraj, and R. S. Moriangthem, "A D-shaped refractive index SPR fiber optic sensor for NIR region," in *SPIE Future Sensing Technologies 2023*, 2023, vol. 12327: SPIE, pp. 293-299.
- [21] L. Singh, Y. Vasimalla, P. Pareek, N. K. Maurya, and R. Kumar, "Utilization of F8BT in refractive index sensor for urea treated urine detection," *Optical and Quantum Electronics*, vol. 55, no. 14, p. 1289, 2023.
- [22] Y. Vasimalla, B. Kaur, S. Maloji, and S. Kumar, "Design and Probing of Prism-Based SPR Nano-Biosensor for Human Sperm Detection," *IEEE Transactions on NanoBioscience*, vol. 24, no. 1, pp. 70-77, 2025, doi: 10.1109/TNB.2024.3419571.

- [23] S. Boddu, Y. Vasimalla, R. Singh, S. Vankayalapati, S. Jain, and S. Kumar, "Label-Free Copper Gallium Sulfide- and Mercury Gallium Sulfide-Based Plasmonic Nanobiosensor for Urea Detection in Urine Samples," *IEEE Transactions on Plasma Science*, pp. 1-8, 2025, doi: 10.1109/TPS.2025.3614683.
- [24] J. Bode, S. Boddu, Y. Vasimalla, and S. Kumar, "A novel gallium phosphide-aluminium antimonide material based SPR sensor with enhanced evanescent field confinement for early detection of cancer biomarkers," *Next Materials*, vol. 9, p. 101222, 2025.
- [25] A. Panda, P. D. Pukhrambam, and G. Keiser, "Performance analysis of graphene-based surface plasmon resonance biosensor for blood glucose and gas detection," *Applied Physics A*, vol. 126, no. 3, p. 153, 2020.
- [26] B. Karki, A. Jha, A. Pal, and V. Srivastava, "Sensitivity enhancement of refractive index-based surface plasmon resonance sensor for glucose detection," *Optical and Quantum Electronics*, vol. 54, no. 9, p. 595, 2022.
- [27] A. Yadav, A. Kumar, and P. Sharan, "Sensitivity enhancement of a plasmonic biosensor for urine glucose detection by employing black phosphorous," *Journal of the Optical Society of America B*, vol. 39, no. 1, pp. 200-206, 2021.
- [28] A. Yadav, S. Kumar, A. Kumar, and P. Sharan, "Effect of 2-D nanomaterials on sensitivity of plasmonic biosensor for efficient urine glucose detection," *Frontiers in Materials*, vol. 9, p. 1106251, 2023.
- [29] S. Kumar, A. Yadav, and B. A. Malomed, "Bimetal thin film, semiconductors, and 2D nanomaterials in SPR biosensors: an approach to enhanced urine glucose sensing," *IEEE Transactions on NanoBioscience*, vol. 23, no. 2, pp. 336-343, 2024.
- [30] B. Karki, A. Uniyal, P. Sarkar, A. Pal, and R. B. Yadav, "Sensitivity improvement of surface plasmon resonance sensor for glucose detection in urine samples using heterogeneous layers: an analytical perspective," *Journal of Optics*, vol. 53, no. 3, pp. 2567-2577, 2024.
- [31] M. El-assar, T. E. Taha, F. E. A. El-Samie, H. A. Fayed, and M. H. Aly, "ZnSe-based highly-sensitive SPR biosensor for detection of different cancer cells and urine glucose levels," *Optical and Quantum Electronics*, vol. 55, no. 1, p. 76, 2023.
- [32] R. R. Rafi, A. S. Mohsin, M. M. Rahman, and M. B. H. Bhuiyan, "Blood Hemoglobin and Urine Glucose Level Detection Using Copper-Graphene Based SPR Biosensor," in *2024 6th International Conference on Electrical Engineering and Information & Communication Technology (ICEEICT)*, 2024: IEEE, pp. 55-60.