

Custom gold-patterned rewritable optical disc based plasmonic sensor for blood hemoglobin detection

HIMADDRI ROY,^{1,†}  MD. EHSANUL KARIM,^{1,2,†}  SUJOY MONDAL,¹
AND SAJID MUHAIMIN CHOUDHURY^{1,*} 

¹Department of Electrical and Electronic Engineering, Bangladesh University of Engineering and Technology, Dhaka- 1205, Bangladesh

²Department of Electrical and Electronic Engineering, BRAC University, Dhaka- 1212, Bangladesh

*sajid@eee.buet.ac.bd

†Co-first authors with equal contributions

Abstract: In this paper, we have presented a novel plasmonic Refractive Index (RI) sensing scheme based on a rewritable optical disc structure with customized dimer-like nanopatterns. We have performed the finite-difference time-domain (FDTD) simulation for this work to exhibit the optical resonant response and sensing performance of our proposed sensing platform. We have also investigated how the changes in the different structural parameters affect the resonant wavelength. The high refractive index sensitivity of 768nm/RIU allowed our sensing system to achieve a resolution limit for blood Hemoglobin(Hb) detection superior to most Hb sensors previously reported. The use of the built-in grating structure of the optical disc, the ability to create customized patterns, and the high resolution achieved in blood Hb detection suggest the potential of our proposed RI sensor as a cost-efficient, reliable sensing technology for medical applications.

© 2022 Optica Publishing Group under the terms of the [Optica Open Access Publishing Agreement](#)

1. Introduction

The utilization of optical sensing structures as biosensors has extensively grown over the recent years due to their expeditious response to the refractive index changes of the medium and super-accurate molecular detection capability. The high sensitivity and label-free sensing of these optical sensors along with electromagnetic interference immunity, and advancements in their on-chip fabrications [1] have escalated their uses in biomolecule detection [2,3], chemical and gas sensing [4], temperature measurement [5], electric field detection [6], environmental monitoring [7] and even in warfare [8]. Now, over these years numerous technological approaches, including photonic crystal [9], ring resonator [10], surface plasmon resonance (SPR) based structure [11], interferometer [12], and evanescent field optical waveguides [13], have been adopted for the realization of optical sensors. Among these mechanisms, SPR-based optical sensing has drawn interest to a greater extent lately from the researchers as an emerging powerful molecular sensing scheme.

Surface plasmon is referred to as electromagnetic waves coupled to charge density oscillations at a metal-dielectric interface resulting in the disappearance of the reflected light under certain conditions causing the phenomenon called resonance. The high sensitivity of the resonance characteristics to the environmental refractive index is exploited in SPR based sensing systems [14]. Besides high sensitivity, provision of multiplexing [15], miniaturization [16], integration with CMOS platforms [17], and remote sensing [18] makes SPR sensing a great alternative to traditional sensing platforms. These advantages make SPR based sensors suitable for applications like bio and chemical sensing [19,20], food and environmental safety [21], imaging [22], drug production [23], and material characterization [24].

Several methods, notably prism and grating coupled and localized SPR, are used for SPR excitation based sensing schemes [25–27]. However, these approaches despite their faster response and high sensitivities, suffer from fabrication complexities and excessive costing problems which make these methodologies incompatible to be manufactured in developing and underdeveloped countries. For instance, the prism-coupled SPR sensors [19,28,29], are unsuitable for miniaturized out-of laboratory applications due to the use of bulky prisms and troublesome angle-control setup [30,31]. Although having emerged as a suitable alternative for compact on-chip sensor applications, grating coupled systems require expensive and complex fabrication processes, for instance, electron-beam lithography [32], nanoimprint [33], interference lithography [34], and photolithography [35]. The invariably high cost and complexity of these fabrication processes somewhat limit the commercialization of SPR based sensing systems. Localized SPR based sensors have gained popularity due to their ability to detect small changes in the surrounding environment [36,37]. However, they too suffer from the same drawback involving sophisticated fabrication requirements [38]. Using these sensor structures may not be acceptable in significant medical applications like hemoglobin concentration measurement in blood or glucose concentration measurement in urine, as the high manufacturing cost and complicated fabrication steps prevent their use commercially on a large scale.

However, a readily available grating surface in the form of a commercial optical disc has recently drawn the attention of researchers as a medium to generate grating-coupled SPR. Using the already in-built grating structures inside these inexpensive discs can eliminate performing all the challenging and sensitive nano-scale patterning tasks required to fabricate the gratings in the previously mentioned grating-coupled sensors. Therefore, the off-the-shelf disc grating is considered a cheap yet efficient alternative as a sensor platform to laboratory fabricated ones. Over the years, Recordable Compact Disc (CD-R) and Digital Versatile Disc (DVD-R) structures have been used in previous literature for sensing applications [39–42]. However, all of these have required some chemical processing to separate the metal grating from other layers. The presence of phase change material (PCM) in DVD Random Access Memory (DVD-RAM) [43] provides additional flexibility to the design, unlike DVD-R, where both the structural and chemical compositions cannot be readily altered. Also, to the best of our knowledge, no sensing system utilizing the commercial DVD-RAM structure has yet been reported.

So, in this article, we have investigated both theoretically and numerically our own designed optical RI sensor based on the DVD-RAM grating structure. The conventional optical switching method used to store and erase data inside a rewritable optical disc allows the patterning of nano-scaled features on the phase change recordable layer of the DVD-RAM in our proposed system. This suggested novel nano-patterning technique can make our fabrication methodology much cheaper than the ones commonly practiced. Our recommended gold dimer patterns increased the bulk refractive index sensitivity through electric field enhancement and provided additional flexibility to our design. The proposed sensor achieved a sensitivity of 768nm/RIU, which is higher than most optical disc-based biosensing systems developed to date. Moreover, we numerically demonstrated its performance as a Hemoglobin concentration level detector, and our design has outperformed the state-of-the-art Hemoglobin concentration measurement techniques in terms of resolution. Considering the possible cost-effectiveness along with the faster and more accurate sensing performance, this study promises its capability as a sensor to be used as an economical and reliable rapid biomolecular (i.e., Hemoglobin, glucose) testing kit for medical applications.

2. Structure layout

Our proposed sensor structure involves an effective modification in the $\text{Ge}_2\text{Sb}_2\text{Te}_5$ (GST) recording layer of the commercially available rewritable disc, i.e., DVD-RAM. This optical disc comprising the top labeled polycarbonate layer, the reflective metal (Aluminum) layer,

and the recording GST layer can provide a cheap grating surface with a grating period (P) of 740 nm. Also, its phase-change recording layer enables scopes for customizations which makes it different from the non-rewritable optical discs, i.e., CD-R, DVD-R and so on. This customization process involves photo-thermal phase-switching that converts the crystalline GST into the amorphous one to obtain our desired pattern. The designed sensing platform as illustrated in Fig. 1, involves a periodic array of dimer-shaped gold patterns deposited in the GST layer. The main objective behind this modification is to intensify the field coupling associated with the gold-GST interface. However, this dimer pattern also offers several adjustable geometric parameters such as gold-dimer length (L), and gold height (H). In section 5, we have rigorously investigated the effects of changes in these parameters on the sensor performance. Hands-on fabrication of the designed sensor is beyond the scope of our work. However, in section 6, we have illustrated our detailed design proposal where each and every step has been mentioned clearly to get to our final sensor design.

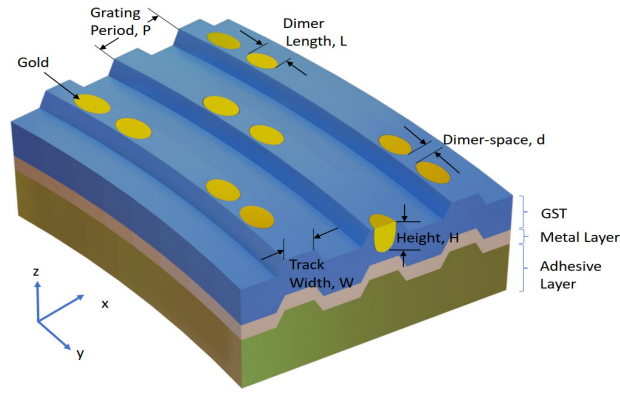


Fig. 1. 3D schematic view of our proposed sensing structure.

We simulated the final three-dimensional (3D) structure utilizing the 3D finite-difference time-domain (FDTD) method for numerical analysis. The unit cell consists of one dimer structure and one period of the grating surface. A x -Plane transverse magnetic (TM) polarized wave is introduced into the cell from the above (in the negative z direction) and a monitor is placed above the source to measure the reflected wave from the dimer. The simulation is periodic in the x and y directions. In order to minimize the memory requirement, we have used anti-symmetric boundary conditions in both x -min and x -max boundaries and symmetric boundary conditions in the y -min and y -max boundaries. A perfectly matched layer (PML) is used in the z -direction. The minimum mesh size is 0.25nm, which is much smaller than the minimum dimension of the structure. We investigated the reflection spectra of this RI sensing model, along with the bulk refractive index (RI) sensitivity of this structure in the results section.

3. Theoretical analysis

The surface plasmon resonance condition involves a coherent fluctuation of free electron charges on the metal boundary. The dispersion relationship relating the frequency and the wave vector of such longitudinal oscillation is given by

$$k_{sp} = \frac{\omega}{c} \sqrt{\frac{\epsilon_m(\omega)n_a^2}{\epsilon_m(\omega) + n_a^2}} \quad (1)$$

where, ω is the angular frequency of oscillation, c is the speed of light in free space, $\epsilon_m(\omega)$ is the gold dielectric function, which can be further split into real (ϵ_{mr}) and imaginary (ϵ_{mi}) parts.

Here, n_a is the refractive index of the analyte. The frequency dependency of the complex metal dielectric function can be formulated by the well-known free electron or Drude model:

$$\epsilon_m(\omega) = \epsilon_{mr}(\omega) + i\epsilon_{mi}(\omega) = 1 - \frac{\Omega_p^2}{\omega^2 - i\omega\Gamma_0} \quad (2)$$

where, Ω_p and Γ_0 are the plasma frequency and damping constant of the metal layer respectively. For gold, $\Omega_p = 1.323 \times 10^{16}$ rad/s and $\Gamma_0 = 1.26 \times 10^{14}$ rad/s [44].

For stable longitudinal electronic oscillation, wave vector matching of the incident light wave and surface plasmon polariton is required. Since the momentum of incident light in air ($k_0 = \omega/c$) is lower than that given by Eq. (1), SPR cannot be excited at a planar dielectric-metal interface. A diffraction grating in optical discs can increase the momentum of an incident optical wave allowing the wave vector matching condition of SPR generation to be fulfilled. The wave vector of this diffracted beam:

$$k_d = n_a \sin\theta + \frac{m2\pi c}{P\omega} \quad (3)$$

where, $m = 0, \pm 1, \pm 2, \dots$ is the diffraction order, $c = 3 \times 10^8$ m/s is the light velocity, and θ is the angle of incidence of the light beam with respect to the grating surface normal (z-axis) and $P = 740\text{nm}$ as illustrated in Fig. 1. The SPR condition for the m^{th} order diffracted beam is given by

$$n_a \sin\theta + \frac{m2\pi c}{P\omega_R} = \text{Re}\{k_{sp}\} \quad (4)$$

where, ω_R is the angular frequency under SPR condition.

For metals like gold exhibiting a high $|\epsilon_{mr}/\epsilon_{mi}|$ ratio, the real part of the SPR propagation constant (k_{sp}) in Eq. (1) may be approximated by [45]

$$\text{Re}\{k_{sp}\} \approx \frac{\omega}{c} \sqrt{\frac{\epsilon_{mr}(\omega)n_a^2}{\epsilon_{mr}(\omega) + n_a^2}}. \quad (5)$$

Combining Eqs. (2)–(5), we have the formulation of the SPR condition for the m^{th} order diffracted beam of a normally incident light ($\theta = 0^\circ$):

$$n_a^2 \omega_R^4 P^2 + [(\Gamma_0^2 - \Omega_p^2)P^2 n_a^2 - 4\pi^2 c^2 m^2 (n_a^2 + 1)]\omega_R^2 - 4\pi^2 c^2 m^2 (\Gamma_0^2 - \Omega_p^2 + n_a^2 \Gamma_0^2) = 0. \quad (6)$$

This equation relates the resonant wavelength (λ_R) and the analyte refractive index for a fixed value of P ($=740\text{nm}$) and m . The solution of Eq. (6) in the near IR region with $n_a=1.3$ and using the relation $\lambda_R = 2\pi c/\omega_R$ gives $\lambda_R = 980.22\text{nm}$ for the 1st order diffracted beam ($m = 1$) of the incident optical wave.

The presence of gold dimer nanoparticles will induce a Localized Surface Plasmon Resonance (LSPR) mode in the spectrum as well. LSPR wavelength can be obtained under the cylindrical dimer approximation using the same methodology as in Ref. [46]. Using values of 170nm, 400nm, and 30nm for the height, diameter, and inter-particle spacing of the cylindrical dimers on these equations yields a value for LSPR wavelength of $\lambda_L = 917.41\text{nm}$.

The particle size and spacing on the surface are expected to have an exponential effect on the LSPR wavelength according to the well-known plasmon ruler equation given by $\Delta\lambda_L/\lambda_L = k \exp(-d/L)$ [46]. The preexponential factor (k) is generally found using curve fitting techniques. This exponential relationship may be approximated to a linear one since $d \ll L$ as illustrated in Fig. 1. Using the Taylor series expansion on this exponential relation, we have

$$\frac{\Delta\lambda_L}{\lambda_L} = k(1 - \frac{d}{2L}). \quad (7)$$

4. Results and discussions

4.1. Reflection spectrum and field profile

Simulating our structure yields a dip in the near-infrared (NIR) region of the reflection spectrum which we can see from Fig. 2. The position of this dip is at a wavelength of approximately 981 nm, which exactly supports the theoretical prediction. The position of the LSPR mode predicted in Section 3, has a slight discrepancy with this dip position due to the cylindrical approximation for the dimer. A closed-form solution for the elliptical hemisphere, which is the exact shape of the dimer, is difficult to obtain using existing theory. A numerical analysis is more suited for such complex structures and is beyond the scope of this work. However, from previous literature, we can intuitively expect the LSPR wavelength to increase as we move from a spherical shape to an elliptical one [47]. This spectrum also presents a Fano resonance-like asymmetric line shape which arises from the coupling between the SPR and LSPR modes arising from the grating structure and the dimer-shaped gold nano-particles respectively [48]. The dark LSPR mode couples with the bright SPR modes through near-field interactions [49].

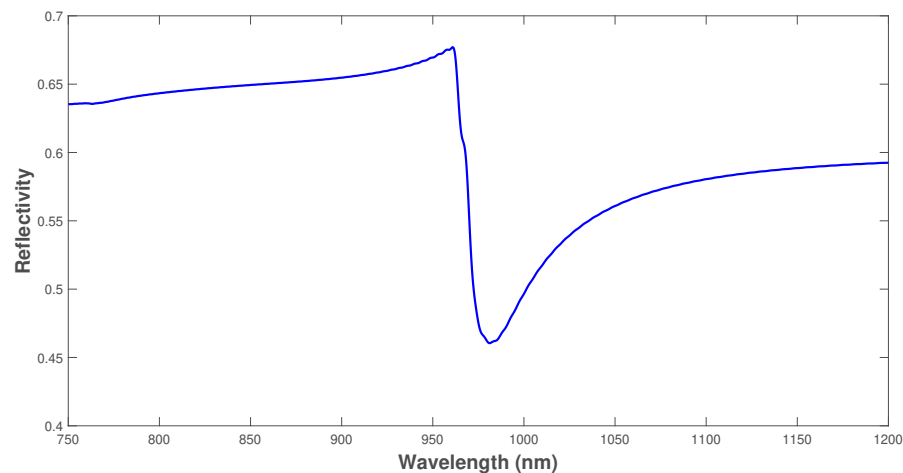


Fig. 2. Reflection spectrum of the proposed DVD-RAM based biosensor structure for standard laser writing dimension of 320nm(*W*)*400nm(*L*), depth, *D*=170nm, and dimer spacing, *d*=30nm. The normally incident TM polarized light source is used for this simulation and the resonance occurs at wavelength of approximately 981 nm.

It becomes evident visually from the XY cross-section electric field profile shown in Fig. 3(a) where a major field confinement is found in the grating slope region. The role of the gold dimer can also be explained from this profile from Fig. 3(a), where we can clearly see an enhanced field distribution around the dimer-surface resulting from field excitation due to surface plasmon-photon coupling in gold. XZ and YZ cross-section profiles in Fig. 3(b) and (c) also support this event of field enhancement due to GCSP resonance.

4.2. Device performance

The performance of optical sensing platforms is generally evaluated in terms of the shift in resonant wavelength ($\Delta\lambda_R$) with changes in the dielectric refractive index (RI) of the surrounding environment (Δn). This is connected to sensitivity defined as $S = \Delta\lambda_R / \Delta n$. In Fig. 4, the resonance dip experiences a redshift from 981.2nm to 1058nm as the bulk refractive index is varied from 1.3 to 1.4. A linear fit to the SPR wavelength shift with changes in RI, as shown in the inset of Fig. 4, translates to a sensitivity of 768nm per RIU. Table 1 compares the RI

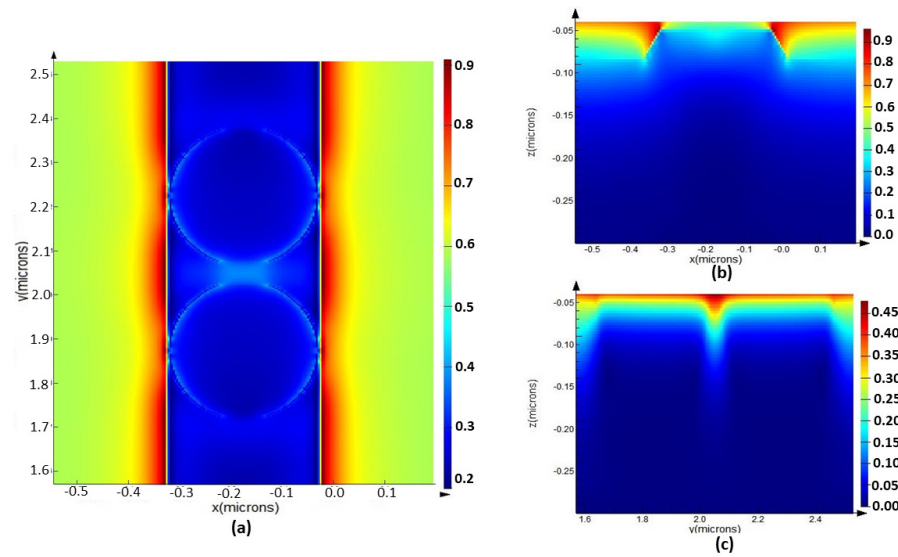


Fig. 3. Normalized electric field profile in the proposed structure with the same configurations as used for Fig. 2. (a) XY cross-section profile presenting the grating-coupled surface plasmon excitation along both sides of the recording track. Also, light-metal coupling in the gold dimer region can be clearly observed in the track. (b) XZ cross-section and (c) YZ cross-section profile both also show the enhanced field characterizations.

sensing performance of our proposed device with other previously reported grating-based sensing platforms including, several structures based on optical discs. The table illustrates a comparable sensing capability of our device to other discussed RI sensors. For completeness, we have also added sensing platforms other than grating-based ones to our comparison. They show quite high sensitivities compared to our structure but require complex fabrication processes or angle control setup, unlike our system. The high sensitivity of our proposed SPR based optical sensor to changes in the surrounding refractive index establishes it as a suitable candidate for biosensing applications. Hemoglobin (Hb) concentration in human blood flow is one such crucial parameter in clinical diagnosis. The normal Hb level for males is 14 to 18 g/dl, and that for females is 12 to 16 g/dl [53]. Abnormal blood Hb concentration, along with other symptoms, can be used to diagnose COPD (Chronic Obstructive Pulmonary Disease), Chronic Lymphocytic Leukemia, Iron deficiency, Polycythemia vera [54–57], to name a few. Long-term persistence of such anomaly in Hb level may even lead to life threatening cardiac, renal, and liver dysfunctions [58]. The variation in the dispersion relationship for solutions with different Hb concentrations is the basis for Hb concentration detection using optical sensing platforms. The FDTD simulation environment requires mapping the Hb concentrations to their respective refractive indices.

The experimental work of Friebel et al. [59] provides a formulation relating the Hb solution concentration and its corresponding refractive index:

$$n(\lambda, c) = n_{H_2O}(\lambda)[\beta(\lambda)c + 1] \quad (8)$$

where, water refractive index (n_{H_2O}) = 1.326 at a reference wavelength of 1000nm [60], $\beta(\lambda) = 2.052 \times 10^{-3}$ dL/g for $\lambda = 1000$ nm [59], c = Hb concentration in g/dL. A linear fit to the data points in Fig. 5 shows a shift in the resonant wavelength of around 2.0885nm for unit g/dL change in hemoglobin levels. Wavelength shifts typically detectable in modern spectrometers are on the order of 0.03nm [61]. For this typical resolving capability of spectrometers, our proposed sensor is expected to detect hemoglobin concentration changes of 0.014 g/dL. The

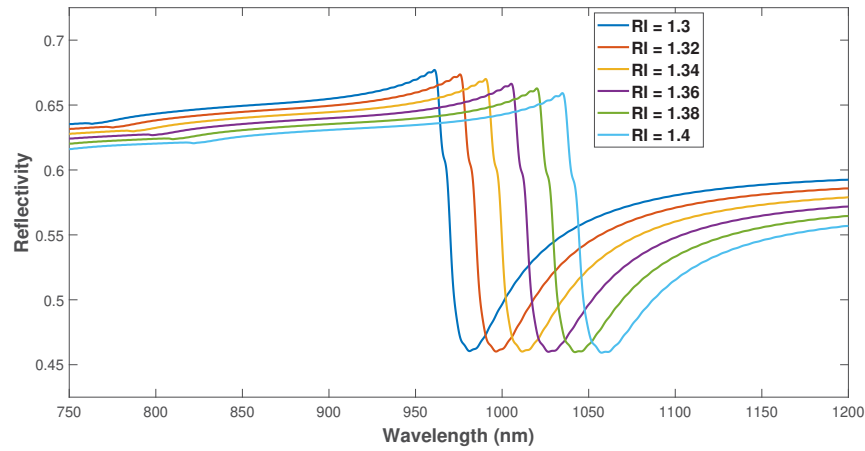


Fig. 4. Plasmonic Fano resonance condition as the environmental refractive index is varied from 1.3 to 1.4 maintaining $W=320\text{nm}$, $L=400\text{nm}$, $H=170\text{nm}$, $d=30\text{nm}$, and the same shape as in Fig. 1. The inset shows the variation of SPR wavelength (λ_R) with changes in bulk refractive index

Table 1. Sensing Performance comparison of our proposed structure with previously reported sensing platforms.

Structure Characteristics	Sensitivity	RI range	Resonance Position	Reference
GST layer on a BaF ₂ grating	1400nm/RIU	1.3-1.34	1033-1039nm	[50]
Au grating on Au,Ti,Si layer	430nm/RIU	1.1-1.4	1600-1700nm	[51]
Isolated metal grating of a DVD-R	637nm/RIU	1.34-1.36	650-700nm	[40]
Multilayer metal on the DVD grating mold	377.7nm/RIU	1.34-1.42	550-600 nm	[42]
Structurally modified DVD-R	850nm/RIU	1.33-1.37	980-1030nm	[39]
Multi-core whispering gallery mode	900nm/RIU	1.33-1.41	1610-1690nm	[52]
Graphene based prism coupled structure	275.15°/RIU	1.338-1.405	589nm	[29]
This work	768nm/RIU	1.3-1.4	981-1058nm	

resolution indicates the lowest amount of change a device can sense, and this value is a significant parameter to evaluate the hemoglobin concentration measurement performance. The lower the resolution is, the more sensitive the sensor is to the changes in Hemoglobin concentration level. Table 2 compares the resolution value of our sensing platform with the existing measurement techniques. Evidently, our designed model provides a better resolution compared to both SPR and non-SPR-based models, which further strengthens its potential to be a revolutionary sensing device for medical applications.

Table 2. Comparison of resolution in g/dL with previously reported hemoglobin detection methods.

	Method	Resolution (g/dL)	Reference
Non-SPR	Multi-wavelength pulse-oximetry based	1.28	[62]
	Spectrophotometry based	1.11	[63]
	BK ₇ /Au/PtSe ₂ /graphene coated sensor	0.6103	[64]
SPR	Au layer based on silica substrate	0.018	[65]
	This work	0.014	

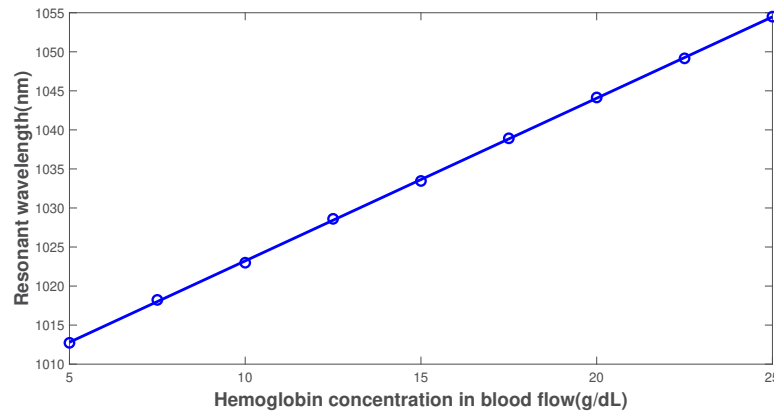


Fig. 5. Resonance behavior of the SPR sensor when exposed to different concentrations of Hb solution. For each of the Hb solutions, standard conditions of $W=320\text{nm}$, $L=400\text{nm}$, $H=170\text{nm}$, $d=30\text{nm}$, and the same etch pit shape as in Fig. 1 was maintained.

5. Effects of structural parameter changes

In this section, we have studied the effects of different structural parameters (e.g. dimer length (L), dimer spacing (d), and dimer shape as depicted in Fig. 1) on the reflection characteristics of the device. These parameters change due to variations in Optical Pickup Units (OPU) of different DVD manufacturers and the external circuitry used to control the laser writing process. So, this study of the effects of these parameters investigates our system's sensitivity to the variation of laser writing systems.

5.1. Dimer length

The dimension of the dimer elements in the y-direction (L) can be controlled by changing the emission time of the laser keeping the write-back speed of the DVD-RAM constant. The emission time depends on the duration of high current input to the laser which can be varied by changing the control bits of commercially available Laser Diode Drivers. In Fig. 6, an increase in L causes a redshift of the SPR wavelength in our proposed system. The electric field, perpendicular to the dimer axis, creates surface charges due to plasma electron confinement in each of the dimer elements. The repulsive dipole-dipole interaction between them, due to the induced surface charges, becomes weaker as L increases, keeping the spacing between them (d) constant. This causes an increase in the resonant wavelength [66]. The plasmon ruler equation described in Eq. (7) also supports this linear trend in the SPR wavelength. Also, the small fractional change in resonant frequency is also validated due to fact that $d < L$.

5.2. Dimer spacing

The two reflection spectra for the two values of dimer spacings (50 and 250nm) in Fig. 7 are almost identical. As the dimensions of dimer elements ($320\text{nm} \times 400\text{nm}$) are much larger than the two spacing values, the shift in resonant wavelength is negligible. This is a trend previously observed with nano-scaled dimers [66]. The small sensitivity of the resonant frequency is also validated by Eq. (7) using this same reasoning. A higher value of d will result in negligible field coupling in the dimers, for which the spacing was limited to 250nm. The dimer spacing depends on the delay between two successive laser emissions, controlled by the timing of control bits of commercial Laser Diode Drivers, as discussed in section 5.1.

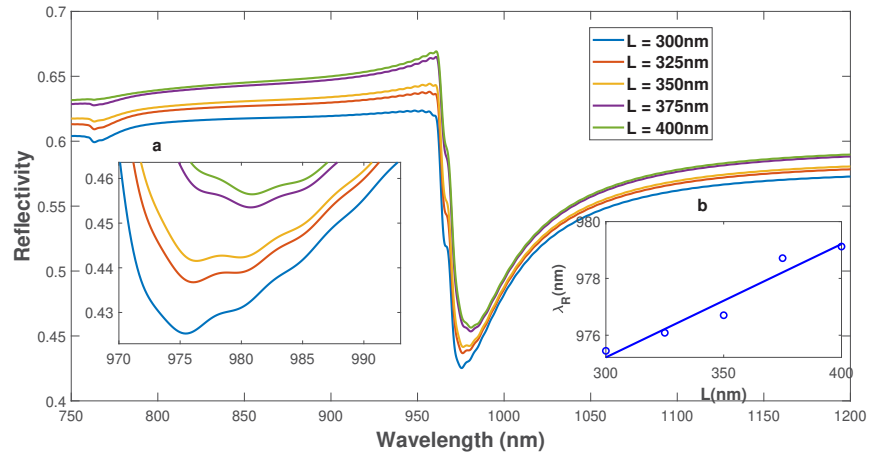


Fig. 6. Resonance condition as the y dimension of each dimer element (L) is varied for a constant bulk RI=1.3, $W=320\text{nm}$, $d=30\text{nm}$, $H=170\text{nm}$ and keeping the shape of the etch pit approximately the same as that in Fig. 1 for all the five values of L . **a** gives a better view of the reflectivity minima. **b** shows the shifting of the resonant wavelength (λ_R) with L .

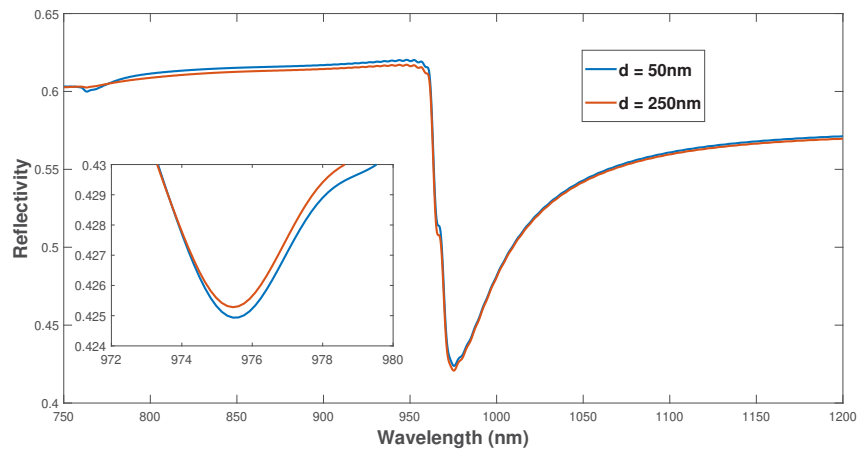


Fig. 7. Plasmonic Fano resonance condition for two different values of dimer spacing (d). For both the values of d , $W=320\text{nm}$, $L=400\text{nm}$, $H=170\text{nm}$, bulk RI=1.3, and the same etch pit shape as in Fig. 1 is maintained. The inset gives a better view of the plasmon dips.

5.3. Etch pit shape

The etch pit shape may change due to an undesirable fluctuation in the laser spot position, caused by diffraction and other non-idealities in the focusing system of the OPU. The inset of Fig. 8 indicates a very negligible shift in the reflectivity minima position as the shape of etch pit in each of the dimer elements is gradually changed from shape I to shape IV in Fig. 8(b-e). This result indicates a negligible sensitivity of the resonance condition to random deflections in the laser spot as long as it is small enough. The widening of dimer element in the y-direction (L) in Fig. 6, and the gradual change of the etch pit shape, from circular to rectangular, in Fig. 8, have the same effect of increasing the amount of gold on the surface. The high extinction coefficient of gold reduces energy coupling to the SPR mode under consideration [67,68]. This results in an upward shift in the reflection spectra in each of the three above-mentioned figures as the amount of gold increases.

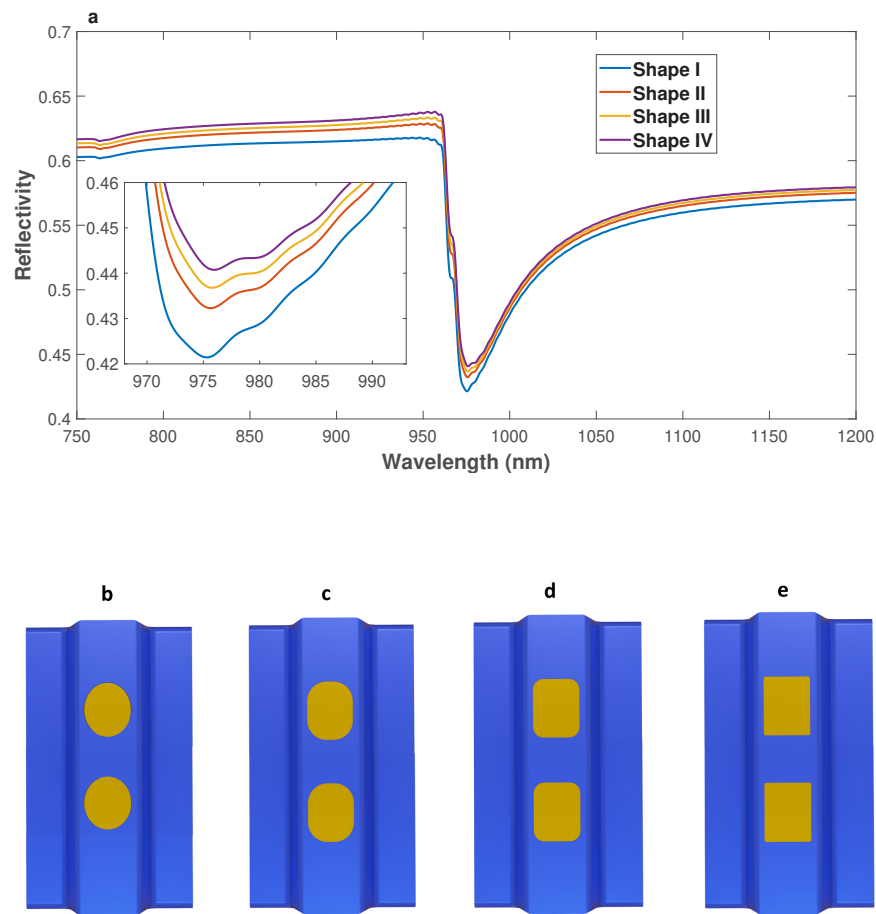


Fig. 8. **a** Changes in the reflection spectrum as the shape of the etch pit is gradually changed from circular (Shape I) to rectangular (Shape IV) for a bulk refractive index of 1.3, $W=320\text{nm}$, $L=400\text{nm}$, $H=170\text{nm}$, $d=30\text{nm}$. The inset shows an enlarged view of the plasmon dips. **b-e** represent the corresponding Shape I to Shape IV respectively.

The small dependence of the resonance condition on the exact location and firing time of the laser beam spot, as studied in Subsections 5.1, 5.2, 5.3 waives the requirement of precise controlling of the laser system. This will allow simpler and cheaper control systems to produce

reliable performance, reducing cost and making the reproducibility of our proposed sensing platform easier to achieve. Also, the heat produced due to SPR generation is not significant enough to make changes in the phase of GST for the continuous wave operation that we are considering in this work [69].

6. Proposed fabrication steps

The required kind of customized shape patterning requires a great amount of control over the optical pickup unit (OPU) inside the DVD-RAM optical drive along with the driver signal power level and pulse-width necessary to record and erase data. There are several open-source software packages available that offer access to the OPU inside the disc drive [70] and enable us to precisely control the laser diode positioning along with the signal power level to have the customized written pattern of our desired size and shape on the recording layer. When the required dimer-shaped amorphous patterning is done, rest of the tentative work-process to build the ultimate tunable biosensor design out of the patterned DVD-RAM has been illustrated stepwise in Fig. 9(a)-(e). To further modify the recording layer with the amorphous dimer patterns written on it as Fig. 9(c), the plastic, i.e., polycarbonate layer adhering to the recording layer is to be peeled off like the way shown in Fig. 9(a)-(b). The next step is to remove the recorded amorphous portions of the GST layer while creating a dimer-shaped array of free space like Fig. 9(d) so that gold can be deposited into those void patterns to fulfill our plan. Here, a good etch selectivity between the crystalline and amorphous states of GST is required to selectively etch out the amorphous while keeping the crystalline parts pretty much intact. In this scenario, Deng et al. [71] proposed tetramethylammonium hydroxide (TMAH) solution as an etchant, having significantly different etching rates for these two states. It is evident from their experimental result that 25% TMAH solution can be an excellent selective wet etchant that can completely wash out the amorphous state from the written patterns without even affecting the crystalline GST state as we expected. Once this etching is done, the etched-away hole areas are filled with gold, and we can finally get the proposed biosensor layout of Fig. 9(e).

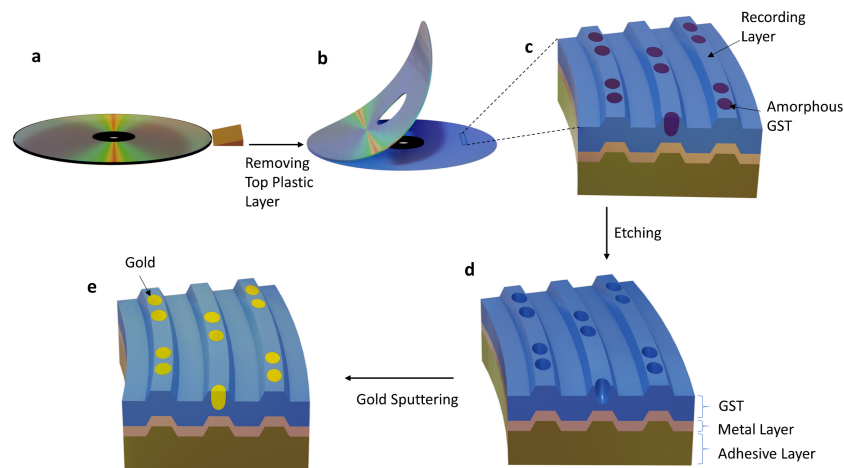


Fig. 9. Proposed tentative step by step work process to obtain the final proposed structure. (a)-(b) The polycarbonate layer attached to the GST being peeled off to get the view of (c), in which the recording layer with a customized data pattern written on it can be observed. (d) The etched out amorphous GST from the GST layer creating hole-array, which is later filled with gold as shown in (e).

7. Conclusion

In this paper, we present a label-free nano-biosensor functioning in the near IR region designed by just modifying the recording layer of a blank rewritable DVD-RAM. The structure includes gold dimer structures in the recording layer of the optical disc for better electric field coupling on the surface. We conducted the theoretical analysis of this design that has validated the position of SPR resonant wavelength found in the numerical analysis. Analyzing the SPR spectra provided a shift of 768nm in resonance dip for a unit change in the bulk refractive index. Additionally, the simulation results show the ability of our proposed plasmonic biosensor to detect Hemoglobin concentration in human blood flow with a precision of 0.014g/dL, which outperforms many of the state-of-the-art detection techniques available right now. We also proposed a cost-efficient and easy fabrication technique for our designed sensor. The proposed surface patterning method includes laser switching of GST and selective etching of its amorphous form as the major steps and has the potential to be used for complex metasurface patterning in future to enhance sensitivity performance [72]. These potentially make our proposed fabrication technique cheaper and less complicated than conventional nanolithography techniques. The findings of our work, considering fabrication feasibility, cost efficiency, and performance of the sensor, show great potential to be used in future biomolecular sensing applications.

Funding. Bangladesh University of Engineering and Technology (CASR Meeting 339, Date 07/04/2021, Resolution 62).

Disclosures. The authors declare no conflict of interest.

Data availability. Data underlying the results presented in this paper are not publicly available at this time but may be obtained from the authors upon reasonable request.

References

1. L. M. Lechuga, "Optical biosensors," *Compr. Analytical Chem.* **44**, 209–250 (2005).
2. P. Damborský, J. Švitel, and J. Katrlík, "Optical biosensors," *Essays Biochem.* **60**(1), 91–100 (2016).
3. M. M. Hassan, F. S. Sium, F. Islam, and S. M. Choudhury, "A review on plasmonic and metamaterial based biosensing platforms for virus detection," *Sens. Bio-Sens. Res.* **33**, 100429 (2021).
4. S. Afsheen, T. Iqbal, M. Aftab, A. Bashir, A. Tehseen, M. Y. Khan, and M. Ijaz, "Modeling of 1d au plasmonic grating as efficient gas sensor," *Mater. Res. Express* **6**(12), 126203 (2019).
5. C. Fenzl, T. Hirsch, and O. S. Wolfbeis, "Photonic crystals for chemical sensing and biosensing," *Angewandte Chemie Int. Ed.* **53**(13), 3318–3335 (2014).
6. V. M. Passaro, P. Barile, and F. De Leonardis, "Investigation of a pin dual-cavity e-field photonic sensor," in *2011 Numerical Simulation of Optoelectronic Devices*, (IEEE, 2011), pp. 59–60.
7. H.-E. Joe, H. Yun, S.-H. Jo, M. B. Jun, and B.-K. Min, "A review on optical fiber sensors for environmental monitoring," *Inter. J. Precision Eng. Manufact. Green Technol.* **5**(1), 173–191 (2018).
8. R. F. Leheny and C. E. McCants, "Technologies for photonic sensor systems," *Proc. IEEE* **97**(6), 957–970 (2009).
9. D. Biallo, M. De Sario, A. D'Orazio, V. Marrocco, V. Petruzzelli, M. Vincenti, F. Prudenzeno, T. Stomeo, M. Grande, G. Visimberga, R. Congolani, and M. de Vittorio, "High sensitivity photonic crystal pressure sensor," *J. Eur. optical. society-Rapid Publ.* **2**, 07017 (2007).
10. L. Ali, M. U. Mohammed, M. Khan, A. H. B. Yousuf, and M. H. Chowdhury, "High-quality optical ring resonator-based biosensor for cancer detection," *IEEE Sens. J.* **20**(4), 1867–1875 (2020).
11. A. Nooke, U. Beck, A. Hertwig, A. Krause, H. Krüger, V. Lohse, D. Negendank, and J. Steinbach, "On the application of gold based spr sensors for the detection of hazardous gases," *Sens. Actuators, B* **149**(1), 194–198 (2010).
12. T. Chalyan, R. Guider, L. Pasquardini, M. Zanetti, F. Falke, E. Schreuder, R. G. Heideman, C. Pederzoli, and L. Pavesi, "Asymmetric mach-zehnder interferometer based biosensors for aflatoxin m1 detection," *Biosensors* **6**(1), 1 (2016).
13. X. Dai, S. J. Mihailov, and C. Blanchetiere, "Optical evanescent field waveguide bragg grating pressure sensor," *Opt. Eng.* **49**(2), 024401 (2010).
14. S. Roh, T. Chung, and B. Lee, "Overview of plasmonic sensors and their design methods," *Proc. SPIE* **7853**, 785303 (2010).
15. F. R. Castiello and M. Tabrizian, "Multiplex surface plasmon resonance imaging-based biosensor for human pancreatic islets hormones quantification," *Anal. Chem.* **90**(5), 3132–3139 (2018).
16. A. Sheridan, P. Ngamukot, P. Bartlett, and J. Wilkinson, "Waveguide surface plasmon resonance sensing: Electrochemical desorption of alkane thiol monolayers," *Sens. Actuators, B* **117**(1), 253–260 (2006).
17. V. Passaro, C. D. Tullio, B. Troia, M. L. Notte, G. Giannoccaro, and F. D. Leonardis, "Recent advances in integrated photonic sensors," *Sensors* **12**(11), 15558–15598 (2012).

18. S. P. Usha, S. K. Mishra, and B. D. Gupta, "Fabrication and characterization of a spr based fiber optic sensor for the detection of chlorine gas using silver and zinc oxide," *Materials* **8**(5), 2204–2216 (2015).
19. P. K. Maharana, R. Jha, and P. Padhy, "On the electric field enhancement and performance of spr gas sensor based on graphene for visible and near infrared," *Sens. Actuators, B* **207**, 117–122 (2015).
20. Z. Sadeghi and H. Shirkani, "High-performance label-free near-infrared spr sensor for wide range of gases and biomolecules based on graphene-gold grating," *Plasmonics* **14**(5), 1179–1188 (2019).
21. J. Homola, J. Dostalek, S. 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).
22. M. Bocková, J. Slabý, T. Špringer, and J. Homola, "Advances in surface plasmon resonance imaging and microscopy and their biological applications," *Annu. Rev. Anal. Chem.* **12**(1), 151–176 (2019).
23. P. Thillaivinayagalingam, J. Gommeaux, M. McLoughlin, D. Collins, and A. R. Newcombe, "Biopharmaceutical production: Applications of surface plasmon resonance biosensors," *J. Chromatogr. B: Anal. Technol. Biomed. Life Sci.* **878**(2), 149–153 (2010).
24. H. Jussila, H. Yang, N. Granqvist, and Z. Sun, "Surface plasmon resonance for characterization of large-area atomic-layer graphene film," *Optica* **3**(2), 151–158 (2016).
25. F.-C. Chien and S.-J. Chen, "A sensitivity comparison of optical biosensors based on four different surface plasmon resonance modes," *Biosens. Bioelectron.* **20**(3), 633–642 (2004).
26. E. Detsi, M. Salverda, P. R. Onck, and J. T. M. De Hosson, "On the localized surface plasmon resonance modes in nanoporous gold films," *J. Appl. Phys.* **115**(4), 044308 (2014).
27. C. Lawrence, N. Geddes, D. Furlong, and J. Sambles, "Surface plasmon resonance studies of immunoreactions utilizing disposable diffraction gratings," *Biosens. Bioelectron.* **11**(4), 389–400 (1996).
28. E. Kretschmann and H. Raether, "Radiative decay of non radiative surface plasmons excited by light," *Zeitschrift für Naturforschung A* **23**(12), 2135–2136 (1968).
29. A. Panda, P. D. Pukhrambam, and G. Keiser, "Performance analysis of graphene-based surface plasmon resonance biosensor for blood glucose and gas detection," *Appl. Phys. A* **126**(3), 153 (2020).
30. J. Cao, Y. Sun, Y. Kong, and W. Qian, "The sensitivity of grating-based spr sensors with wavelength interrogation," *Sensors* **19**(2), 405 (2019).
31. K. Gürel, "Coupled surface plasmon structures and applications," Ph.D. thesis, bilkent university (2009).
32. X. Hoa, A. Kirk, and M. Tabrizian, "Enhanced spr response from patterned immobilization of surface bioreceptors on nano-gratings," *Biosens. Bioelectron.* **24**(10), 3043–3048 (2009).
33. J. Chen, J. Shi, D. Decanini, E. Cambril, Y. Chen, and A.-M. Haghiri-Gosnet, "Gold nanohole arrays for biochemical sensing fabricated by soft uv nanoimprint lithography," *Microelectron. Eng.* **86**(4-6), 632–635 (2009).
34. D. Kotlarek, M. Vorobii, W. Ogieglo, W. Knoll, C. Rodriguez-Emmenegger, and J. Dostálek, "Compact grating-coupled biosensor for the analysis of thrombin," *ACS Sens.* **4**(8), 2109–2116 (2019).
35. S. Long, J. Cao, Y. Wang, S. Gao, N. Xu, J. Gao, and W. Wan, "Grating coupled spr sensors using off the shelf compact discs and sensitivity dependence on grating period," *Sens. Actuators Rep.* **2**(1), 100016 (2020).
36. Z. Guo, X. Yang, F. Shen, Q. Zhou, J. Gao, and K. Guo, "Active-tuning and polarization-independent absorber and sensor in the infrared region based on the phase change material of Ge₂Sb₂Te₅ (gst)," *Sci. Rep.* **8**(1), 12433 (2018).
37. L. Shengyong, A. Xiaochuan, W. Ronghua, and C. Jiajun, "Design and simulation verification an environmental change metamaterial sensor," *Opt. Commun.* **428**, 251–257 (2018).
38. B. Sepúlveda, P. C. Angelomé, L. M. Lechuga, and L. M. Liz-Marzán, "Lspr-based nanobiosensors," *Nano Today* **4**(3), 244–251 (2009).
39. X. Dou, P.-Y. Chung, P. Jiang, and J. Dai, "Surface plasmon resonance and surface-enhanced raman scattering sensing enabled by digital versatile discs," *Appl. Phys. Lett.* **100**(4), 041116 (2012).
40. Y. Sun, S. Sun, M. Wu, S. Gao, and J. Cao, "Refractive index sensing using the metal layer in dvd-r discs," *RSC Adv.* **8**(48), 27423–27428 (2018).
41. T. Arai, S. C. Gopinath, H. Mizuno, P. K. Kumar, C. Rockstuhl, K. Awazu, and J. Tominaga, "Toward biological diagnosis system based on digital versatile disc technology," *Jpn. J. Appl. Phys.* **46**(6B), 4003–4006 (2007).
42. R. Ahmed, M. O. Ozen, M. G. Karaaslan, C. A. Prator, C. Thanh, S. Kumar, L. Torres, N. Iyer, S. Munter, S. Southern, T. J. Henrich, F. Inci, and U. Demirci, "Tunable fano-resonant metasurfaces on a disposable plastic-template for multimodal and multiplex biosensing," *Adv. Mater.* **32**(19), 1907160 (2020).
43. M. H. Lankhorst, B. W. Ketelaars, and R. A. Wolters, "Low-cost and nanoscale non-volatile memory concept for future silicon chips," *Nat. Mater.* **4**(4), 347–352 (2005).
44. L. J. Mendoza Herrera, D. M. Arboleda, D. C. Schinca, and L. B. Scaffardi, "Determination of plasma frequency, damping constant, and size distribution from the complex dielectric function of noble metal nanoparticles," *J. Appl. Phys.* **116**(23), 233105 (2014).
45. J. Homola, I. Koudela, and S. S. Yee, "Surface plasmon resonance sensors based on diffraction gratings and prism couplers: sensitivity comparison," *Sens. Actuators, B* **54**(1-2), 16–24 (1999).
46. P. K. Jain, W. Huang, and M. A. El-Sayed, "On the universal scaling behavior of the distance decay of plasmon coupling in metal nanoparticle pairs: a plasmon ruler equation," *Nano Lett.* **7**(7), 2080–2088 (2007).
47. P. K. Jain and M. A. El-Sayed, "Surface plasmon resonance sensitivity of metal nanostructures: physical basis and universal scaling in metal nanoshells," *J. Phys. Chem. C* **111**(47), 17451–17454 (2007).

48. Z. Qi, J. Yao, L. Zhao, Y. Cui, and C. Lu, "Tunable double-resonance dimer structure for surface-enhanced raman scattering substrate in near-infrared region," *Photonics Res.* **3**(6), 313–316 (2015).
49. H. Liu, B. Zhao, X. Zhang, H. Zhang, B. Wu, and S. Tang, "High-sensitive numerical gas detection using lspr effect and fano resonance in a slotted mdm structure," *Photonic Sens.* **12**(2), 164–174 (2022).
50. X. Zou, G. Zheng, Y. Chen, F. Xian, and L. Xu, "Actively tunable mid-infrared fano resonance in $\text{Ge}_2\text{Sb}_2\text{Te}_5$ -based grating structures," *Opt. Mater.* **88**, 54–59 (2019).
51. K. Q. Le, Q. M. Ngo, and T. K. Nguyen, "Nanostructured metal–insulator–metal metamaterials for refractive index biosensing applications: design, fabrication, and characterization," *IEEE J. Sel. Top. Quantum Electron.* **23**(2), 388–393 (2017).
52. K. Chakrabarti, M. S. Obaidat, S. Mostufa, and A. K. Paul, "Design and analysis of a multi-core whispering gallery mode bio-sensor for detecting cancer cells and diabetes tear cells," *OSA Continuum* **4**(8), 2294–2307 (2021).
53. H. H. Billett, "Hemoglobin and hematocrit," *Clinical Methods: The History, Physical, and Laboratory Examinations*. 3rd edition (1990).
54. N. Milman, "Anemia—still a major health problem in many parts of the world!" *Ann. Hematol.* **90**(4), 369–377 (2011).
55. M. Hallek, B. D. Cheson, D. Catovsky, F. Caligaris-Cappio, G. Dighiero, H. Döhner, P. Hillmen, M. J. Keating, E. Montserrat, K. R. Rai, and T. J. Kipps, "Guidelines for the diagnosis and treatment of chronic lymphocytic leukemia: a report from the international workshop on chronic lymphocytic leukemia updating the national cancer institute–working group 1996 guidelines," *Blood, The J. Am. Soc. Hematol.* **111**(12), 5446–5456 (2008).
56. C. Cote, M. Zilberberg, S. Mody, L. Dordelly, and B. Celli, "Haemoglobin level and its clinical impact in a cohort of patients with copd," *Eur. Respir. J.* **29**(5), 923–929 (2007).
57. B. J. Stuart and A. J. Viera, "Polycythemia vera," *American family physician* **69**, 2139–2144 (2004).
58. S. Allali, V. Brousse, A.-S. Sacri, M. Chalumeau, and M. de Montalembert, "Anemia in children: prevalence, causes, diagnostic work-up, and long-term consequences," *Expert Rev. Hematology* **10**(11), 1023–1028 (2017).
59. M. Friebe and M. Meinke, "Model function to calculate the refractive index of native hemoglobin in the wavelength range of 250–1100 nm dependent on concentration," *Appl. Opt.* **45**(12), 2838–2842 (2006).
60. A. N. Bashkatov and E. A. Genina, "Water refractive index in dependence on temperature and wavelength: a simple approximation," *Proc. SPIE* **5068**, 393–395 (2003).
61. Z. Yang, L. Xia, S. Li, R. Qi, X. Chen, and W. Li, "Highly sensitive refractive index detection based on compact hsc-spr structure in a microfluidic chip," *Sens. Actuators, A* **297**, 111558 (2019).
62. M. R. Macknet, S. Norton, P. L. Kimball-Jones, R. L. Applegate, R. D. Martin, and M. W. Allard, "Noninvasive measurement of continuous hemoglobin concentration via pulse co-oximetry," *Chest* **132**(4), 493C (2007).
63. L. Lamhaut, R. Apriotesei, M. Lejay, B. Vivien, and P. Carli, "Comparison between a new non-invasive continuous technology of spectrophotometry-based and rbc count for haemoglobin monitoring during surgery with hemorrhagic risk: 3ap7–1," *Eur. J. Anaesthesiol.* **27**, 60–61 (2010).
64. 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* **4**(8), 2164–2176 (2021).
65. A. K. Sharma, "Plasmonic biosensor for detection of hemoglobin concentration in human blood: Design considerations," *J. Appl. Phys.* **114**(4), 044701 (2013).
66. W. Rechberger, A. Hohenau, A. Leitner, J. Krenn, B. Lamprecht, and F. Aussenegg, "Optical properties of two interacting gold nanoparticles," *Opt. Commun.* **220**(1–3), 137–141 (2003).
67. C. J. Alleyne, A. G. Kirk, R. C. McPhedran, N.-A. P. Nicorovici, and D. Maystre, "Enhanced spr sensitivity using periodic metallic structures," *Opt. Express* **15**(13), 8163–8169 (2007).
68. K. M. Byun, S. J. Kim, and D. Kim, "Grating-coupled transmission-type surface plasmon resonance sensors based on dielectric and metallic gratings," *Appl. Opt.* **46**(23), 5703–5708 (2007).
69. A. Sierant, R. Panaś, J. Fiutowski, H.-G. Rubahn, and T. Kawalec, "Tailoring optical discs for surface plasmon polaritons generation," *Nanotechnology* **31**(2), 025303 (2020).
70. E. E.-T. Hwu and A. Boisen, "Hacking cd/dvd/blu-ray for biosensing," *ACS Sens.* **3**(7), 1222–1232 (2018).
71. C. Deng, Y. Geng, and Y. Wu, "Selective wet etching of $\text{Ge}_2\text{Sb}_2\text{Te}_5$ phase-change thin films in thermal lithography with tetramethylammonium," *Appl. Phys. A* **104**(4), 1091–1097 (2011).
72. K. Chakrabarti, S. Mostufa, and A. K. Paul, "Design and analysis of a position chirped metamaterial photonic crystal array for confinement of light pulses," *J. Opt.* **23**(11), 115201 (2021).