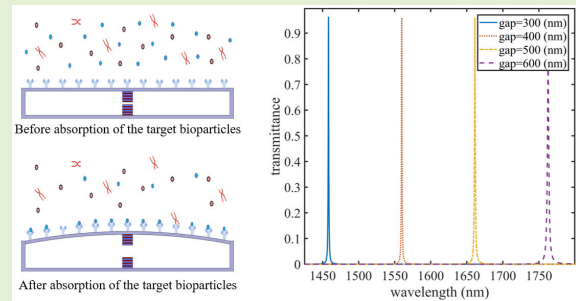


An Ultrasensitive MOEMS Platform Relied on an Adjustable Defect Width for Label-Free Bioparticle Detection

Yashar Gholami, Behnam Saghirzadeh Darki, Mohammad Hossein Poorghadiri Isfahani, Kian Jafari^{ID}, *Senior Member, IEEE*, and Taha Azad

Abstract—This article presents a novel optomechanical surface stress-based biosensor. The proposed device employs a functionalized microelectromechanical systems (MEMS) transducer to capture the target bioparticles in the sample. The interaction of the bioreceptors immobilized on the transducer surface with the target bioparticles modifies the transducer's surface stress, which results in its deformation. The accurate measurement of the transducer deformation allows for label-free detection of the target bioparticles and their concentration. In order to measure the transducer deformation, an optical system consisting of a defective 1-D photonic crystal with adjustable defect width is utilized, which significantly enhances the sensor sensitivity. This BioMEMS platform exhibits improved functional characteristics, including a bandwidth (BW) of 5 kHz, full-width at half-maximum (FWHM) of 0.8 nm, quality factor of 1822.5, mechanical sensitivity of $13.5 \mu\text{m}/\text{Nm}^{-1}$, optical sensitivity of 1.02, and total sensitivity of $13.775 \mu\text{m}/\text{Nm}^{-1}$. These functional characteristics demonstrate the capability of the proposed sensor to detect various diseases in their early stages or progression, which is highly desirable in point-of-care testing (POCT) applications.

Index Terms—BioMEMS, optical biosensing, photonic crystal, point-of-care testing (POCT), surface stress.



I. INTRODUCTION

PPOINT-OF-CARE-TESTING (POCT) refers to clinical laboratory testing that is conducted in close proximity to the site of patient care, where treatment is provided. POCT enables rapid turnaround of test results and has the potential to generate a result quickly, allowing for prompt implementation of appropriate treatment. As a result, POCT can lead to improved clinical and economic outcomes when compared to traditional laboratory testing [1], [2].

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The conventional laboratory testing approach entails a complex series of steps, starting from collecting patient samples, transporting them to a centralized laboratory that could be located far away, and subjecting them to multiple processing procedures. This process can lead to delays in treatment and impede prompt clinical decision-making. However, POCT offers a solution to this challenge by bringing the laboratory to the patient [3], [4].

The advancements in electronics and instrumentation have contributed to the development of more precise and compact POCT devices [5], [6]. The microelectromechanical systems (MEMS) technology, which has already proven its capabilities in various fields, such as inertial sensors [7], [8], logic circuits [9], [10], and built-in self-tests methods [11], [12], have shown great potential in POCT applications [13]. This technology offers several advantages in the medical field and POCT devices, such as small size, low cost, scalability, high accuracy, and sensitivity [14]. As a result, the MEMS technology can be utilized to create a wide range of biosensors capable of detecting various diseases and their progression.

Additionally, MEMS-based biosensors provide label-free detection methods that eliminate the need for secondary binding assays. This results in reduced cost and time for the diagnosis process, as there are no additional preparation

steps required [15], [16]. To achieve this, a MEMS transducer converts the quantity of target molecules into a measurable mechanical property. MEMS transducers can operate in either static or dynamic mode. In the dynamic mode, the changes in the mechanical resonant frequency are detected as the mass of target bioparticles on the sensor surface increases. The magnitude of the resonant frequency variation can determine the type, quantity, or concentration of the target molecules [17]. To achieve high mass sensitivity, two key factors are necessary: devices with nanoscale dimensions and high-quality factors (1000–100 000) that require measurements in vacuum. However, for sensing biomolecules, measurements should be carried out in aqueous solutions, the natural environment for biological processes to occur. Nanomechanical resonators in liquids exhibit a very low-quality factor (1)–(10) due to viscous damping. Even with miniaturization down to the nanoscale, these limitations cannot be improved [18].

On the other hand, biosensors based on static mode measure the concentration of target bioparticles in a gas or aqueous solution by analyzing the steady-state deformation of the MEMS transducer. This deformation is caused by surface stress resulting from the interaction between the transducer and target bioparticles [19], [20]. The measurement of displacements can be achieved through optical [21] and electrical techniques [22], [23]. However, utilizing electrical detection methods, such as piezoresistive and capacitive detection, can result in increased complexity during fabrication and limitations on sensor geometry and materials. On the other hand, the optical methods allow for higher sensitivity and more flexibility in the fabrication of nanomechanical systems. This flexibility provides many opportunities for the development of innovative concepts [20], [24].

One of the earliest optical methods utilized for detecting target bioparticles involves using the position sensing (photograph) method, which is employed to measure the displacement of the MEMS transducer. It is highly versatile as it can be used in vacuum, air, and liquids. This method involves measuring the deflection of a laser beam reflected off the sensor surface using a position-sensitive photodetector. Despite its efficacy, this readout technique faces challenges when used in high-density sensor arrays, due to the intricate optical alignment required for the light source, sensing device, and detectors [25], [26].

To address these challenges, new methods relying on Fabry–Perot interferometers have been introduced, in which the displacement of the MEMS transducer is determined by measuring the intensity of light at a specific wavelength [21], [27]. However, they have a major limitation, which is the need for a stable signal source, since their intensity may vary as a function of time [28].

Furthermore, extensive efforts have been made to develop biosensors with all-optical detection methods based on the changes in the refractive index, such as surface plasmon resonance (SPR) [29], [30], metamaterials [31], [32], and photonic crystal-based approaches [33]. In these approaches, alterations in the concentration of the target bioparticles lead to modifications in the refractive index of the sample and consequently cause variations in the transmission or reflection

spectrum of the device. Nevertheless, the minimal variation in the refractive index in response to slight changes in the concentration of the target particles in the sample presents an ongoing challenge that needs to be addressed.

On the other hand, the mechanical displacement of MEMS transducers of the surface stress-based biosensors induces substantial changes in optical properties, including alterations in light intensity and wavelength spectrum shift. Consequently, integrating MEMS transducers into biosensor structures represents an effective method for the detection of biomarkers at minimal concentrations.

In order to tackle the aforementioned challenges, this article presents an innovative surface-stress-based BioMEMS sensor. The proposed platform utilizes a functionalized MEMS transducer and a defective 1-D photonic crystal with an adjustable defect width. For an optical measurement, the defect in the photonic crystal structure creates a resonant peak within its bandgap. As the target molecules in the sample are captured by the bioreceptors immobilized on the surface of the MEMS transducer, the surface stress on the transducer changes, leading to its deformation and consequently altering the defect's width in the photonic crystal. This change in the defect's width causes a shift in the resonant wavelength of the defective photonic crystal, allowing for the measurement of the concentration of target bioparticles in the sample. Due to its compact size and high sensitivity, this innovative device is well-suited for a variety of POCT applications.

The rest of this article is organized as follows. Section II presents the operation principles and mechanical and optical analyses of the proposed device. Finally, the functional characteristics of the present BioMEMS platform compared to several related contributions are discussed in Section III.

II. DESIGN AND OPERATIONAL PRINCIPLES

The design of the proposed surface stress-based BioMEMS sensor is illustrated in Fig. 1. This device incorporates a parylene-C membrane as the MEMS transducer and a defective 1-D photonic crystal for the measurement of the membrane deformation. Photonic crystals are multilayer structures, in which the refractive index of the constituent materials varies periodically. These crystals are characterized by their photonic bandgaps resulting from multiple Bragg scattering of light at the interface between two different media. The central wavelength of the band gap (λ_c) can be determined as follows [34]:

$$\lambda_c = \frac{2(n_1 \times a + n_2 \times b)}{m}, \quad m = 1, 2, 3, \dots \quad (1)$$

where n_1 and n_2 represent the refractive indices of the materials in the photonic crystal, while a and b denote the thickness of the first and second periodic layers of the 1-D photonic crystal. For any 1-D photonic crystal consisting of two materials with refractive indices n_1 and n_2 and thicknesses a and b , respectively, the maximum bandgap's width occurs when $n_1 a = n_2 b$ with the central wavelength of $\lambda_c = 4n_1 a = 4n_2 b$ [35]. Furthermore, introducing a defect gap in photonic crystals creates a single transmission peak within the bandgap, also known as the resonant mode. This peak occurs as a result of a disruption in the periodicity of the structure layers.

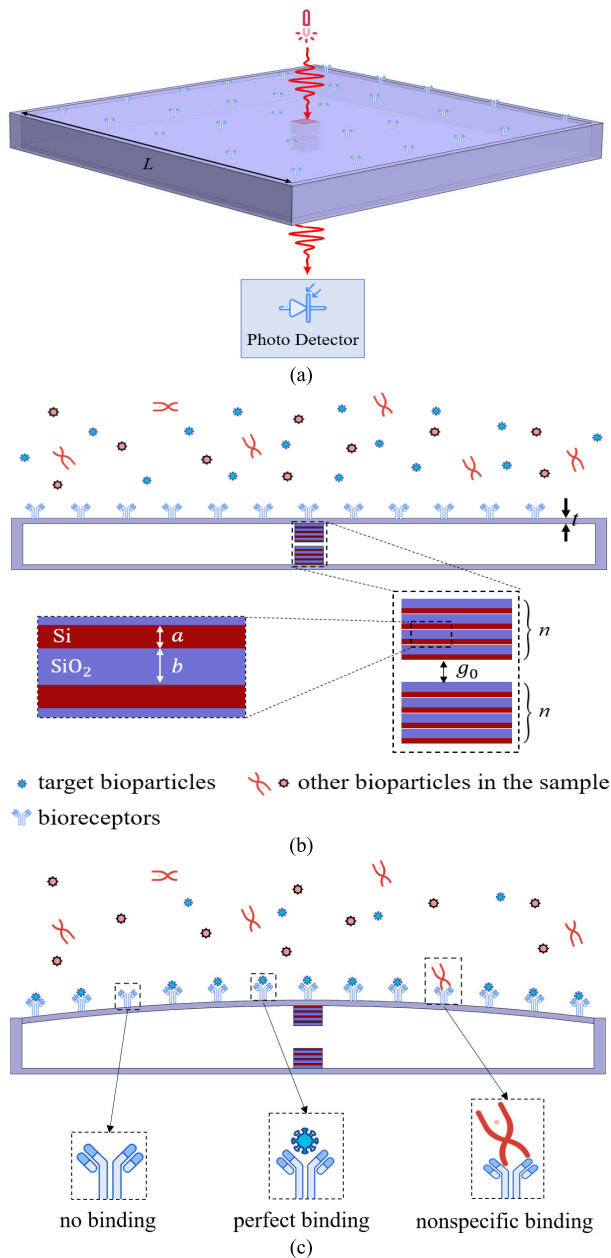


Fig. 1. (a) Design of the proposed surface-stress-based BioMEMS sensor. (b) Schematic of the immobilized receptors on the surface of the proposed structure. (c) Deformation of the membrane under compressive surface stress.

In this design, the photonic crystal has been divided into two parts by a defect featuring a primary defect width of g_0 . The lower part of the photonic crystal is fixed, while the upper part is suspended by the membrane, which can move upward or downward with the deformation of the membrane.

As shown in Fig. 1(b), the biological receptors are immobilized on the membrane's surface. When the target bioparticles in the sample bind to these receptors, chemical interactions create surface stress on the surface of the membrane. The change in surface stress leads to membrane deformation. The curvature of the membrane (κ) caused by a surface stress change can be calculated using Stoney equation [36]

$$\kappa = \frac{6(1-\nu)}{Et^2} \Delta\sigma \quad (2)$$

where E , ν , t , and $\Delta\sigma$ represent Young's modulus, Poisson's ratio, membrane thickness, and the change in surface stress, respectively. Additionally, the vertical displacement of the membrane's center can be determined as follows:

$$\Delta z = \frac{\kappa a^2}{C} \quad (3)$$

where a denotes the membrane's side length, and C is a constant that depends on the boundary conditions and the specific shape of the membrane deformation. For a square membrane with fixed edges, the constant C is approximately $C \approx 2\pi^2$. By combining (3) and (4), we can derive the expression for the vertical displacement of the membrane's center

$$\Delta z = \frac{6(1-\nu)}{CEt^2} \Delta\sigma a^2. \quad (4)$$

This indicates that the vertical displacement at the center of the membrane increases with an increase in the side length of the membrane.

It is notable that each bioreceptor is designed to specifically capture a target bioparticle and is theoretically not supposed to interact with other molecules in the test sample. However, in practical applications, it is important to consider the potential for nonspecific binding, as there exists a small probability of binding between bioreceptors and other bioparticles in the sample (nonspecific binding). In addition, after the bioreceptors are immobilized, it is necessary to cover the sensing areas with block reagents, such as bovine serum albumin, casein, or gelatin to prevent nonspecific binding on the sensing surface. Additionally, increasing the density of bioreceptors on the sensing surface not only enhances the sensor's sensitivity but also minimizes the probability of nonspecific interactions by reducing the free surface area available for nontarget molecules. This ensures the selectivity and reliability of the proposed platform in detecting target bioparticles [18]. Finally, the membrane's deformation alters the defect width of the photonic crystal, resulting in a shift in the resonant mode wavelength. This shift is indicative of the concentration of the target biomolecules. The dimensions and physical parameters of the proposed biosensor are detailed in Table I.

A. Mechanical Analysis

As previously mentioned, the photonic crystal's defect width variation is achieved through the use of the MEMS transducer. This transducer consists of a 350-nm-thick parylene-C membrane with immobilized bioreceptors on its surface. Upon the absorption of target molecules by the receptors, the membrane undergoes deformation as described by (2). To model the surface stress, a compressive force is applied to the upper side edges of the membrane [23], [37]. As shown in Fig. 2, the absorption of target molecules and the alteration of surface stress result in the greatest displacement at the center of the membrane.

Fig. 3 illustrates the displacement of the membrane's center with various surface stress values for different side lengths of the membrane. The results reveal that larger side lengths correspond to greater displacement of the membrane. The displacement, caused by the application of surface stress,

TABLE I
DIMENSIONS AND PHYSICAL PARAMETERS OF THE PROPOSED
BIOMEMS SENSOR

Symbol	Description	Value
E	Young's module of parylene-C	4.5 GPa
ρ	Density of parylene-C	1.289 g/cm ³
ν	Poisson's ratio of parylene-C	0.4
ϵ_{par-C}	Dielectric constant of parylene-C	3
ϵ_{Si}	Dielectric constant of Si	11.7
ϵ_{SiO2}	Dielectric constant of SiO ₂	3.9
t	Thickness of membrane	0.35 μ m
L	Side length of membrane	200 μ m
g_0	Initial width of defect	350 nm
a	Thickness of Si layers	110 nm
b	Thickness of SiO ₂ layers	270 nm
n	Number of pair-layers at each side of the defect	4

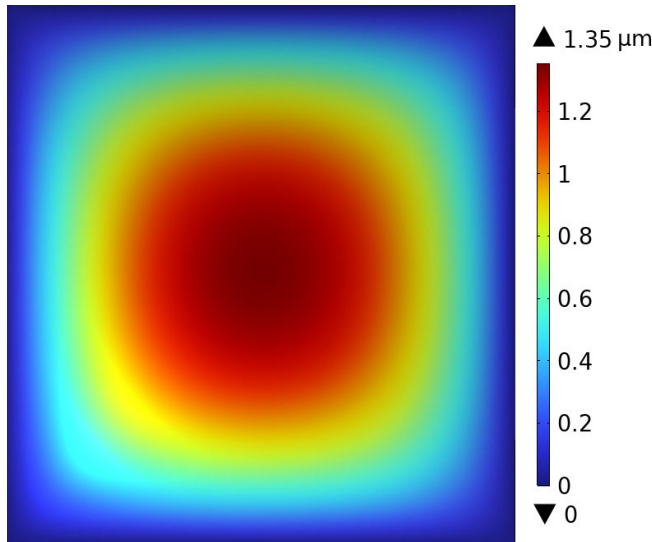


Fig. 2. Contour plot showing the vertical displacement of the membrane under compressive surface stress, simulated using COMSOL Multiphysics.

determines the mechanical sensitivity of the sensor according to

$$S_m = \frac{\Delta z}{\Delta \sigma} \quad (5)$$

where Δz is the displacement of the center of the membrane and $\Delta \sigma$ is the surface stress variation on the top surface of the membrane due to the absorption of target bioparticles. Hence, the longer the membrane's side length, the greater its mechanical sensitivity. Nevertheless, using larger dimensions for the membrane due to its thinness can lead to challenges during its release. Therefore, for this design, we have chosen a membrane side length of 200 μ m. Based on these dimensions, the mechanical sensitivity of the proposed biosensor is 13.5 μ m/Nm⁻¹.

Furthermore, the frequency response of BioMEMS sensors plays a significant role in determining the device's bandwidth

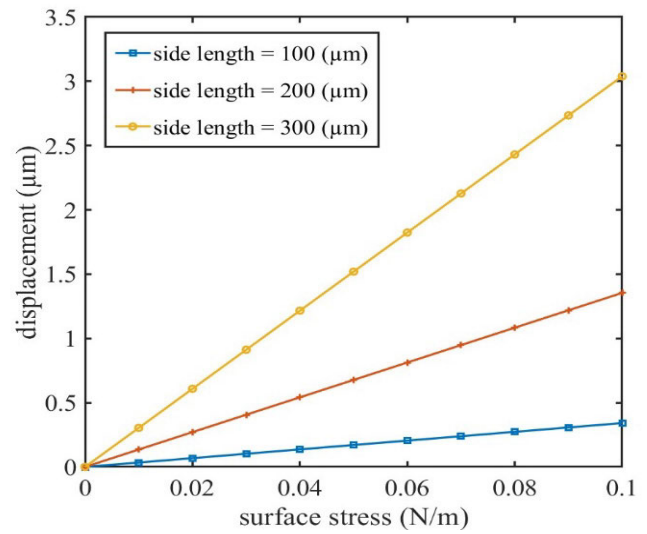


Fig. 3. Displacement of the center of the membrane with the variation in surface stress.

(BW). The resonant frequencies of a membrane with fixed boundaries are determined by solving the wave equation under appropriate boundary conditions. For a square-shaped membrane with the side length of a , the natural frequencies are given by

$$f_{mn} = \frac{1}{2\pi} \sqrt{\frac{T}{\rho h} \left(\frac{m^2 \pi^2}{a^2} \right) + \left(\frac{n^2 \pi^2}{a^2} \right)} \quad (6)$$

where T is the membrane tension per unit length, ρh is the areal mass density, and mn are mode indices. The first resonant frequency f_{11} corresponds to the lowest mode ($m = 1, n = 1$) and is approximated as follows:

$$f_{11} = \frac{1}{2\pi} \sqrt{\frac{12T}{\rho h a^3}} \quad (7)$$

If the membrane is modeled as a spring-mass system with an effective spring constant k and mass m , the first resonance frequency simplifies to

$$f_{11} = \frac{1}{2\pi} \sqrt{\frac{k}{m}} \quad (8)$$

where $k \approx 4T/a$ and $m \approx (1/3)\rho h a^2$. Thus, the primary resonant frequency of the present device can be defined as

$$f_r = \frac{1}{2\pi} \sqrt{\frac{k}{m}} 23.3 \text{ kHz} \quad (9)$$

The first six mechanical resonant modes of the proposed biosensor are illustrated in Fig. 4. In order to determine the BW of the proposed sensor, the frequency response is also shown in Fig. 5. The results obtained through numerical methods confirm those obtained through analytical approaches. As depicted in the figure, the frequency response curve remains entirely flat across a broad range of frequencies preceding the first resonant mode. Consequently, one can determine the proposed biosensor's BW as BW = 5 kHz.

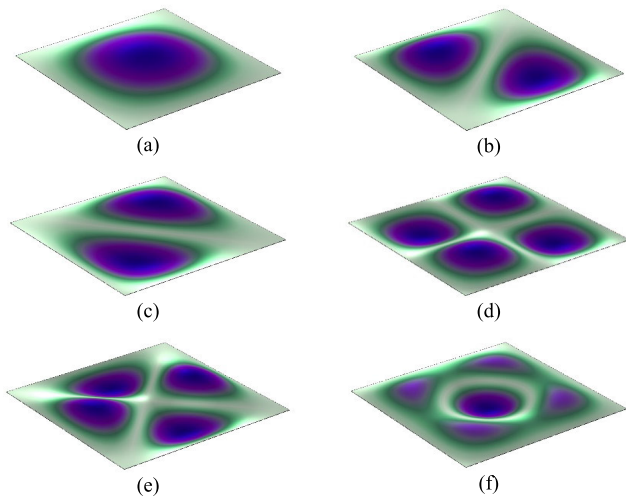


Fig. 4. First six resonant modes of the proposed BioMEMS sensor. (a) f_1 : 23.3 kHz, (b) f_2 : 46.1 kHz, (c) f_3 : 46.1 kHz, (d) f_4 : 74.2 kHz, (e) f_5 : 87.2 kHz, and (f) f_6 : 92.1 kHz.

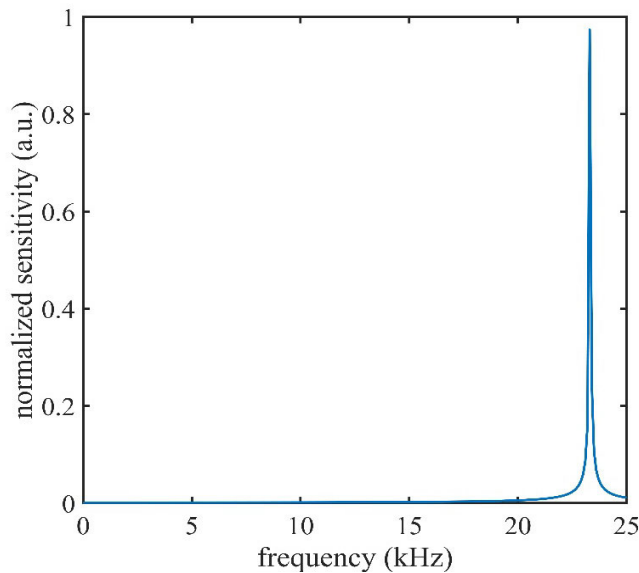


Fig. 5. Frequency response of the present device.

B. Optical Analysis

The optical detection system of the proposed sensor incorporates a 1-D defective photonic crystal composed of periodic layers of Si and SiO₂. The transfer matrix method (TMM) has been employed as the theoretical model for calculating the transmittance spectrum of the proposed photonic crystal structure. The theoretical transmittance results derived from TMM were cross-validated with numerical simulations conducted in COMSOL Multiphysics, demonstrating strong alignment and confirming the accuracy of our approach. Based on (1), the central bandgap wavelength of this photonic crystal is 1550 nm. The presence of a defect within the photonic crystal structure induces a resonant mode within the bandgap. As indicated in Fig. 6, the peak wavelength of the resonant mode is 1458 nm.

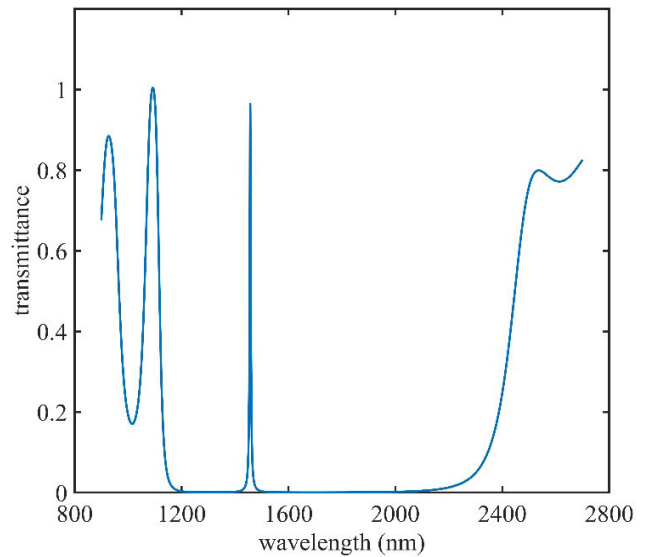


Fig. 6. Transmission spectrum of the proposed BioMEMS sensor with the defect width of 300 nm.

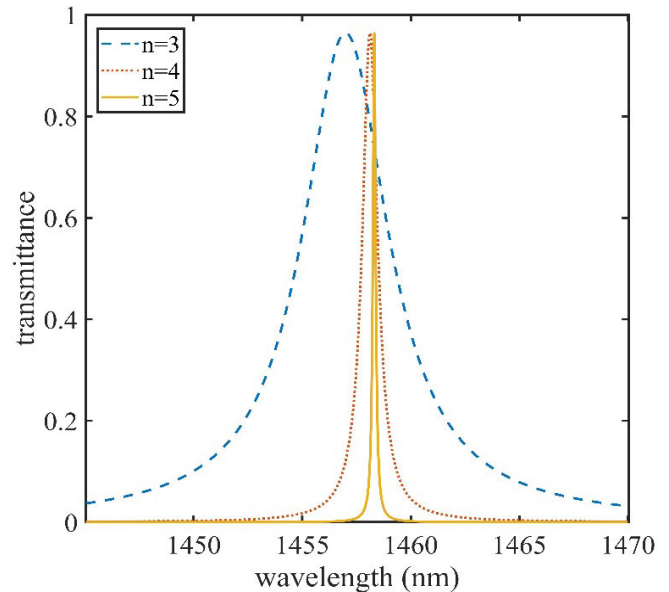


Fig. 7. Variation in resonant mode width with the number of pair layers at each side of the photonic crystal defect.

It is notable that the width of the resonant mode decreases as the number of pair layers at each side of the defect of the photonic crystal increases (Fig. 7). In the proposed design, due to the complexity of fabrication for the higher number of pair layers, each part of the photonic crystal is designed with four pair layers, resulting in a fullwidth at halfmaximum (FWHM) of 0.8 nm. Accordingly, the quality factor (Q) of the proposed device can be determined as follows:

$$Q = \frac{\lambda_r}{\text{FWHM}} = \frac{1458}{0.8} = 1822.5 \quad (10)$$

where λ_r is the resonant mode's wavelength

The absorption of target molecules in the sample causes the membrane to compress, increasing the width of the defect (Δg) in the photonic crystal. This deformation leads to a

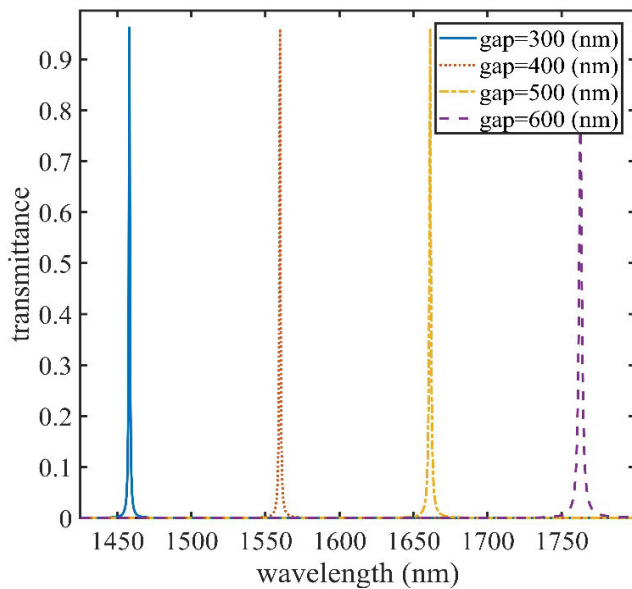


Fig. 8. Wavelength shift of the resonant mode in the transmission spectrum of the proposed biosensor with different defect widths in the photonic crystal structure.

shift in the wavelength of the resonant mode (Fig. 8), which determines the absorption rate of the target bioparticles in the sample. The optical sensitivity of the sensor can be calculated according to

$$S_o = \frac{\Delta\lambda}{\Delta z} = 1.02 \quad (11)$$

where $\Delta\lambda$ is the wavelength shift of the resonant mode peak of the defective photonic crystal.

Considering the mechanical and optical sensitivity of the proposed BioMEMS sensor, the total sensitivity of this device is finally obtained as follows:

$$S = \frac{\Delta\lambda}{\Delta\sigma} = \frac{\Delta z}{\Delta\sigma} \times \frac{\Delta\lambda}{\Delta z} = 13.77 \frac{\mu\text{m}}{\text{Nm}^{-1}}. \quad (12)$$

III. COMPARATIVE STUDY

The performance of the proposed biosensor was assessed by comparing its functional characteristics with those of other similar devices utilizing optical detection methods. The results of the comparison are outlined in Table II. It should be mentioned that some of the compared contributions utilize all-optical approaches [29], [38], while the others are surface stress-based biosensors employing MEMS transducers [39], [40] to detect the type and concentration of the bioparticles. The all-optical biosensors detect changes in the sample's refractive index. However, since the changes in the refractive index can be influenced by factors other than the concentration of the target molecules, these methods exhibit lower selectivity compared to surface stress-based biosensors. Consequently, sensors based on surface stress exhibit more efficiency in the early diagnosis and monitoring of various diseases.

The proposed biosensor demonstrates significantly reduced FWHM due to the use of a defective photonic crystal in the optical detection system. This results in a substantially higher quality factor compared to other contributions based on

TABLE II
FUNCTIONAL CHARACTERISTICS OF THE PROPOSED BioMEMS SENSOR COMPARED TO SEVERAL RECENT CONTRIBUTIONS

	Functional mechanism	FWHM (nm)	Q	S_o	S_m ($\mu\text{m}/\text{Nm}^{-1}$)	S ($\mu\text{m}/\text{Nm}^{-1}$)
[38]	All optical Metamaterial	316	6.1	-	-	-
[29]	All optical Plasmonic	53	23.77	-	-	-
[39]	MEMS Photonic-crystal	1.7	647	0.28	4.52	1.26
[40]	MEMS Metamaterial	20	102.7	0.7	11.55	8.08
proposed device	MEMS Photonic-crystal	0.8	1822.5	1.02	13.5	13.77

MEMS technology. The utilization of a movable membrane made of parylene-C with low Young's modulus has led to greater displacement, increasing the mechanical sensitivity of the device. Additionally, varying the defect width causes a significant shift in the resonant mode wavelength, enhancing the optical sensitivity of the device. With its high mechanical and optical sensitivity, the proposed sensor is expected to exhibit greater total sensitivity than other works, a notion confirmed by the results in Table II. Considering these functional characteristics, one can conclude that the proposed BioMEMS platform outperforms similar contributions in overall performance, making it an appealing option for POCT applications.

IV. CONCLUSION

This article introduces a novel surface-stress-based BioMEMS sensor. This sensor leverages an optical detection system using a defective 1-D photonic crystal with a variable defect width. The utilization of this optical method significantly enhances the sensor's sensitivity, enabling label-free detection of the type and concentration of the target bioparticles. The BioMEMS device exhibits impressive functional characteristics, including a BW of 5 kHz, an FWHM of 0.8 nm, a quality factor of 1822, an optical sensitivity of 1.02, a mechanical sensitivity of $13.5 \mu\text{m}/\text{Nm}^{-1}$, and a total sensitivity of $13.77 \mu\text{m}/\text{Nm}^{-1}$. Based on these functional characteristics, the proposed platform is well-suited for the detection of various bioparticles at very low concentrations, which is crucial for early disease diagnosis and shows great potential for point-of-care testing applications.

REFERENCES

- [1] M. Zhang, X. Cui, and N. Li, "Smartphone-based mobile biosensors for the point-of-care testing of human metabolites," *Mater. Today Bio*, vol. 14, Mar. 2022, Art. no. 100254, doi: 10.1016/j.mtbio.2022.100254.
- [2] T. Yan, G. Zhang, H. Chai, L. Qu, and X. Zhang, "Flexible biosensors based on colorimetry, fluorescence, and electrochemistry for point-of-care testing," *Frontiers Bioeng. Biotechnol.*, vol. 9, Sep. 2021, Art. no. 753692, doi: 10.3389/fbioe.2021.753692.
- [3] Ö. Erdem, I. Eş, G. A. Akceoglu, Y. Saylan, and F. Inci, "Recent advances in microneedle-based sensors for sampling, diagnosis and monitoring of chronic diseases," *Biosensors*, vol. 11, no. 9, p. 296, Aug. 2021, doi: 10.3390/bios11090296.

- [4] R. Park, S. Jeon, J. Jeong, S.-Y. Park, D.-W. Han, and S. W. Hong, "Recent advances of point-of-care devices integrated with molecularly imprinted polymers-based biosensors: From biomolecule sensing design to intraoral fluid testing," *Biosensors*, vol. 12, no. 3, p. 136, Feb. 2022, doi: [10.3390/bios12030136](https://doi.org/10.3390/bios12030136).
- [5] S. Singh, P. S. Podder, M. Russo, C. Henry, and S. Cinti, "Tailored point-of-care biosensors for liquid biopsy in the field of oncology," *Lab Chip*, vol. 23, no. 1, pp. 44–61, 2023, doi: [10.1039/d2lc00666a](https://doi.org/10.1039/d2lc00666a).
- [6] S. Suan Ng, H. Ling Lee, P. Bothi Raja, and R.-A. Doong, "Recent advances in nanomaterial-based optical biosensors as potential point-of-care testing (PoCT) probes in carcinoembryonic antigen detection," *Chem.-Asian J.*, vol. 17, no. 14, Jul. 2022, Art. no. 202200287, doi: [10.1002/asia.202200287](https://doi.org/10.1002/asia.202200287).
- [7] S. Shi et al., "High sensitivity MEMS accelerometer using PZT-based four L-shaped beam structure," *IEEE Sensors J.*, vol. 22, no. 8, pp. 7627–7636, Apr. 2022, doi: [10.1109/JSEN.2022.3155407](https://doi.org/10.1109/JSEN.2022.3155407).
- [8] A. K. N. Shotorban, K. Jafari, and K. Abedi, "Optical MEMS accelerometer sensor relying on a micro-ring resonator and an elliptical disk," *IET Circuits, Devices Syst.*, vol. 13, no. 7, pp. 1102–1106, Oct. 2019, doi: [10.1049/iet-cds.2019.0029](https://doi.org/10.1049/iet-cds.2019.0029).
- [9] Y. Gholami, F. Marvi, R. Ghorbanloo, M. R. Eslami, and K. Jafari, "High speed binary neural network hardware accelerator relied on optical NEMS," *IEEE Trans. Nanotechnol.*, vol. 23, pp. 63–69, 2024, doi: [10.1109/TNANO.2023.3343618](https://doi.org/10.1109/TNANO.2023.3343618).
- [10] Y. Gholami, A. Sheikholeslami, F. Marvi, K. Jafari, and M. H. Moaiyeri, "Design and analysis of optical nano electro-mechanical-system-based logic circuits relied on movable waveguides," *Opt. Eng.*, vol. 61, no. 10, Oct. 2022, Art. no. 107110, doi: [10.1117/1.OE.61.10.107110](https://doi.org/10.1117/1.OE.61.10.107110).
- [11] K. Jafari, "A parameter estimation approach based on binary measurements using maximum likelihood analysis—Application to MEMS," *Int. J. Control, Autom. Syst.*, vol. 15, no. 2, pp. 716–721, Apr. 2017, doi: [10.1007/s12555-015-0343-1](https://doi.org/10.1007/s12555-015-0343-1).
- [12] O. Legendreet et al., "A low-cost built-in self-test method for thermally actuated resistive MEMS sensors," *Sens. Actuators A, Phys.*, vol. 194, pp. 8–15, May 2013, doi: [10.1016/j.sna.2013.01.013](https://doi.org/10.1016/j.sna.2013.01.013).
- [13] F. Marvi, K. Jafari, M. Shahabadi, M. Tabaradz, F. Ghorbani-Bidkorpeh, and T. Azad, "Ultrasensitive detection of vital biomolecules based on a multi-purpose BioMEMS for point of care testing: Digoxin measurement as a case study," *Sci. Rep.*, vol. 14, no. 1, p. 1633, Jan. 2024, doi: [10.1038/s41598-024-51864-4](https://doi.org/10.1038/s41598-024-51864-4).
- [14] A. Gupta, D. Rotake, and A. Darji, "Ultrasensitive detection of lead ions through portable sensor technology for enhanced environmental monitoring," *IEEE Sensors Lett.*, vol. 8, no. 7, pp. 1–4, Jul. 2024, doi: [10.1109/lsens.2024.3409363](https://doi.org/10.1109/lsens.2024.3409363).
- [15] F. Marvi and K. Jafari, "A measurement platform for label-free detection of biomolecules based on a novel optical BioMEMS sensor," *IEEE Trans. Instrum. Meas.*, vol. 70, pp. 1–7, 2021, doi: [10.1109/TIM.2021.3052001](https://doi.org/10.1109/TIM.2021.3052001).
- [16] R. M. R. Pinto, V. Chu, and J. P. Conde, "Label-free biosensing of DNA in microfluidics using amorphous silicon capacitive micro-cantilevers," *IEEE Sensors J.*, vol. 20, no. 16, pp. 9018–9028, Aug. 2020, doi: [10.1109/JSEN.2020.2986497](https://doi.org/10.1109/JSEN.2020.2986497).
- [17] F. Tehrani et al., "An integrated wearable microneedle array for the continuous monitoring of multiple biomarkers in interstitial fluid," *Nat. Biomed. Eng.*, vol. 6, pp. 1214–1224, May 2022, doi: [10.1038/s41551-022-00887-1](https://doi.org/10.1038/s41551-022-00887-1).
- [18] J. Tamayo, P. M. Kosaka, J. J. Ruz, Á. S. Paulo, and M. Calleja, "Biosensors based on nanomechanical systems," *Chem. Soc. Rev.*, vol. 42, no. 3, pp. 1287–1311, 2013, doi: [10.1039/C2CS35293A](https://doi.org/10.1039/C2CS35293A).
- [19] R. Mathew and A. R. Sankar, "A review on surface stress-based miniaturized piezoresistive SU-8 polymeric cantilever sensors," *Nano-Micro Lett.*, vol. 10, no. 2, pp. 1–41, Feb. 2018, doi: [10.1007/s40820-018-0189-1](https://doi.org/10.1007/s40820-018-0189-1).
- [20] M. Khorsandifard, K. Jafari, and A. Sheikholeslami, "A proposal for a novel surface-stress based BioMEMS sensor using an optical sensing system for highly sensitive diagnoses of bio-particles," *Sens. Imag.*, vol. 22, no. 1, pp. 1–16, Jul. 2021, doi: [10.1007/s11220-021-00355-1](https://doi.org/10.1007/s11220-021-00355-1).
- [21] T. Takahashi, T. Hizawa, N. Misawa, M. Taki, K. Sawada, and K. Takahashi, "Surface stress sensor based on MEMS Fabry-Pérot interferometer with high wavelength selectivity for label-free biosensing," *J. Micromech. Microeng.*, vol. 28, no. 5, May 2018, Art. no. 054002, doi: [10.1088/1361-6439/aaa80](https://doi.org/10.1088/1361-6439/aaa80).
- [22] N. Hernández-Sebastián, D. Díaz-Alonso, F. J. Renero-Carrillo, N. Villa-Villaseñor, and W. Calleja-Arriaga, "Design and simulation of an integrated wireless capacitive sensors array for measuring ventricular pressure," *Sensors*, vol. 18, no. 9, p. 2781, Aug. 2018, doi: [10.3390/s18092781](https://doi.org/10.3390/s18092781).
- [23] M. Z. Ansari, M. Bisen, and C. Cho, "Modelling and analysis of diaphragm integrated SU8/CB nanocomposite piezoresistive polymer microcantilever biosensor," *Microsyst. Technol.*, vol. 24, no. 7, pp. 3061–3068, Jul. 2018, doi: [10.1007/s00542-018-3777-6](https://doi.org/10.1007/s00542-018-3777-6).
- [24] N. Maheshwari, G. Chatterjee, and V. R. Rao, "A technology overview and applications of bio-MEMS," *J. ISSS*, vol. 3, no. 2, pp. 39–59, Jan. 2014.
- [25] L. Y. Beaulieu, M. Godin, O. Laroche, V. Tabard-Cossa, and P. Grütter, "A complete analysis of the laser beam deflection systems used in cantilever-based systems," *Ultramicroscopy*, vol. 107, nos. 4–5, pp. 422–430, 2007, doi: [10.1016/j.ultramicro.2006.11.001](https://doi.org/10.1016/j.ultramicro.2006.11.001).
- [26] S. F. Memon, M. M. Ali, J. T. Pembroke, B. S. Chowdhry, and E. Lewis, "Measurement of ultralow level bioethanol concentration for production using evanescent wave based optical fiber sensor," *IEEE Trans. Instrum. Meas.*, vol. 67, no. 4, pp. 780–788, Apr. 2018, doi: [10.1109/TIM.2017.2761618](https://doi.org/10.1109/TIM.2017.2761618).
- [27] K. Takahashi et al., "Elastomer-based MEMS optical interferometric transducers for highly sensitive surface stress sensing for biomolecular detection," *MRS Commun.*, vol. 9, no. 1, pp. 381–389, Mar. 2019, doi: [10.1557/mrc.2019.11](https://doi.org/10.1557/mrc.2019.11).
- [28] A. B. Socorro-Leránz, D. Santano, I. Del Villar, and I. R. Matias, "Trends in the design of wavelength-based optical fibre biosensors (2008–2018)," *Biosensors Bioelectron.*, X, vol. 1, Jun. 2019, Art. no. 100015, doi: [10.1016/j.biosx.2019.100015](https://doi.org/10.1016/j.biosx.2019.100015).
- [29] R. Yan et al., "Highly sensitive plasmonic nanorod hyperbolic metamaterial biosensor," *Photon. Res.*, vol. 10, no. 1, pp. 84–95, 2022, doi: [10.1364/PRJ.444490](https://doi.org/10.1364/PRJ.444490).
- [30] C. Li, J. Gao, and M. Shafi, "Optical fiber SPR biosensor complying with a 3D composite hyperbolic metamaterial and a graphene film," *Photon. Res.*, vol. 9, no. 3, pp. 379–388, 2021, doi: [10.1364/PRJ.416815](https://doi.org/10.1364/PRJ.416815).
- [31] M. Y. Azab, M. F. O. Hameed, A. M. Nasr, and S. S. A. Obayya, "Highly sensitive metamaterial biosensor for cancer early detection," *IEEE Sensors J.*, vol. 21, no. 6, pp. 7748–7755, Mar. 2021, doi: [10.1109/JSEN.2021.3051075](https://doi.org/10.1109/JSEN.2021.3051075).
- [32] W. Fang, Z. Ma, X. Lv, J. Liu, W. Pei, and Z. Geng, "Flexible terahertz metamaterial biosensor for label-free sensing of serum tumor marker modified on a non-metal area," *Opt. Exp.*, vol. 30, no. 10, p. 16630, 2022, doi: [10.1364/oe.454647](https://doi.org/10.1364/oe.454647).
- [33] A. Sinibaldi, "Cancer biomarker detection with photonic crystals-based biosensors: An overview," *J. Lightw. Technol.*, vol. 39, no. 12, pp. 3871–3881, 2021, doi: [10.1109/JLT.2021.3056225](https://doi.org/10.1109/JLT.2021.3056225).
- [34] J. Zhao, X. Li, L. Zhong, and G. Chen, "Calculation of photonic band-gap of one dimensional photonic crystal," *J. Phys., Conf. Ser.*, vol. 183, Aug. 2009, Art. no. 012018, doi: [10.1088/1742-6596/183/1/012018](https://doi.org/10.1088/1742-6596/183/1/012018).
- [35] S. Elshahat, I. Abood, M. S. M. Esmail, Z. Ouyang, and C. Lu, "One-dimensional topological photonic crystal mirror heterostructure for sensing," *Nanomaterials*, vol. 11, no. 8, p. 1940, Jul. 2021, doi: [10.3390/nano11081940](https://doi.org/10.3390/nano11081940).
- [36] W. Zhang et al., "Structural optimization of the micro-membrane for a novel surface stress-based capacitive biosensor," *Microelectron. Eng.*, vol. 106, pp. 9–12, Jun. 2013, doi: [10.1016/j.mee.2013.01.040](https://doi.org/10.1016/j.mee.2013.01.040).
- [37] G. Yoshikawa, T. Akiyama, S. Gautsch, P. Vettiger, and H. Rohrer, "Nanomechanical membrane-type surface stress sensor," *Nano Lett.*, vol. 11, no. 3, pp. 1044–1048, Mar. 2011, doi: [10.1021/nl103901a](https://doi.org/10.1021/nl103901a).
- [38] S. Seyyedmasoumian, A. Attariabad, A. Pourziad, and M. Bemani, "Refractive index biosensor using metamaterial perfect absorber based on graphene in near-infrared for disease diagnosis," *IEEE Sensors J.*, vol. 22, no. 15, pp. 14870–14877, Aug. 2022, doi: [10.1109/JSEN.2022.3187011](https://doi.org/10.1109/JSEN.2022.3187011).
- [39] F. Marvi and K. Jafari, "An ultra-sensitive 1D photonic crystal BioNEMS platform for label free detection of biomolecules," *IEEE Sensors J.*, vol. 21, no. 20, pp. 22927–22933, Oct. 2021, doi: [10.1109/JSEN.2021.3108000](https://doi.org/10.1109/JSEN.2021.3108000).
- [40] F. Marvi, K. Jafari, and M. Sawan, "Grating bio-microelectromechanical platform architecture for multiple biomarker detection," *Biosensors*, vol. 14, no. 8, p. 385, Aug. 2024, doi: [10.3390/bios14080385](https://doi.org/10.3390/bios14080385).



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