

An Ultrasensitive BioMEMS Sensor Based on the Phase Modulation Optical Systems

Capteur BioMEMS ultrasensible basé sur les systèmes optiques à modulation de phase

Yashar Gholami^{ID}, Zahra Alinia, Behnam Saghirzadeh Darki^{ID}, Kian Jafari^{ID}, *Senior Member, IEEE*,
and Mohammad Hossein Moaiyeri^{ID}, *Senior Member, IEEE*

Abstract—This article presents an ultrasensitive surface stress-based BioMEMS platform with an optical biosensing detection method. The proposed biosensor consists of two main parts: a microelectromechanical systems (MEMS) transducer, which converts the chemical interaction of the bioreceptors with the target bioparticles into mechanical displacement, and an optical system to detect the displacement of the MEMS transducer and determine the concentration of the target bioparticles. This design uses a membrane held by six stands above a waveguide as the MEMS transducer to capture the target bioparticles in the test sample. The absorption of the target bioparticles by the bioreceptors, which are immobilized on the surface of the movable membrane, creates surface stress on the top surface of the membrane, leading to its deformation. While the movable part approaches the waveguide, it interacts with the modes' evanescent field, increasing the effective refractive index. Finally, the refractive index variation causes a shift in the mode's phase that determines the concentration of the target bioparticles. The operational characteristics of the present biosensor resulting from numerical and analytical approaches are as follows: phase shift of 250π , optical sensitivity of 1935π rad/RIU, mechanical sensitivity of $1.64 \mu\text{m}/\text{N}\cdot\text{m}^{-1}$, and figure of merit (FOM) of $1.29 \pi\text{rad}/\text{RIU}\mu\text{m}$. The obtained results indicate that the proposed biosensor has the potential to be employed in point-of-care (POC) tests. This would enable the detection of target biomolecules associated with specific diseases and the measurement of their concentrations, which is indicative of disease progression.

Résumé—Cet article présente une plateforme BioMEMS ultrasensible basée sur les contraintes de surface et dotée d'une méthode de détection par biodétection optique. Le biocapteur proposé se compose de deux parties principales : un transducteur microélectromécanique (MEMS), qui convertit l'interaction chimique des biorécepteurs avec les bioparticules cibles en déplacement mécanique, et un système optique permettant de détecter le déplacement du transducteur MEMS et de déterminer la concentration des bioparticules cibles. Cette conception utilise une membrane maintenue par six supports au-dessus d'un guide d'ondes comme transducteur MEMS pour capturer les bioparticules cibles dans l'échantillon testé. L'absorption des bioparticules cibles par les biorécepteurs, qui sont immobilisés à la surface de la membrane mobile, crée une contrainte de surface sur la surface supérieure de la membrane, entraînant sa déformation. Lorsque la partie mobile s'approche du guide d'ondes, elle interagit avec le champ évanescent des modes, augmentant ainsi l'indice de réfraction effectif. Enfin, la variation de l'indice de réfraction provoque un décalage de la phase du mode qui détermine la concentration des bioparticules cibles. Les caractéristiques opérationnelles du présent biocapteur résultant d'approches numériques et analytiques sont les suivantes : déphasage de 250π , sensibilité optique de 1935π rad/RIU, sensibilité mécanique de $1,64 \mu\text{m}/\text{N}\cdot\text{m}^{-1}$ et facteur de qualité (FOM) de $1,29 \pi$ rad/RIU μm . Les résultats obtenus indiquent que le biocapteur proposé pourrait être utilisé dans les tests au point de service (POC). Cela permettrait de détecter les biomolécules cibles associées à des maladies spécifiques et de mesurer leurs concentrations, ce qui est révélateur de la progression de la maladie.

Index Terms—BioMEMS, optical biosensing, phase shift, point-of-care (POC) testing.

Received 17 January 2025; revised 11 May 2025 and 22 August 2025; accepted 1 September 2025. Date of publication 4 November 2025; date of current version 10 November 2025. (Corresponding author: Kian Jafari.)

Yashar Gholami, Zahra Alinia, and Mohammad Hossein Moaiyeri are with the Faculty of Electrical Engineering, Shahid Beheshti University, Tehran 1983969411, Iran.

Behnam Saghirzadeh Darki is with the Department of Electrical Engineering and the Digital Processing and Machine Vision Research Center, Na.C., Islamic Azad University, Najafabad 8514143131, Iran.

Kian Jafari is with the Université de Sherbrooke Cancer Research Institute (IRCUS), the Faculty of Engineering and the Laboratoire Nanotechnologies Nanosystèmes (LN2) - CNRS UMI-3463, Université de Sherbrooke, Sherbrooke, QC J1N 3C6, Canada (e-mail: kian.jafari@usherbrooke.ca).

Associate Editor managing this article's review: Stefano Gregori.

Digital Object Identifier 10.1109/ICJECE.2025.3608553

I. INTRODUCTION

OVER the past decade, significant advancements have been observed in microelectromechanical systems (MEMS) technology. The ease of fabrication, coupled with high reliability, compact dimensions, advanced manufacturing processes, and cost-effectiveness, has facilitated the widespread application of MEMS in various fields, including inertial sensors [1], [2], [3], built-in self-test methods [4], [5], logic circuits, and processors [6].

Recently, MEMS technology demonstrated high capabilities in the detection and characterization of diverse biological

particles, such as cellular constituents [7], [8], viruses [9], and proteins [10]. These devices have numerous applications in different areas, including healthcare [11], environmental monitoring [12], and food safety [13]. Furthermore, bio-MEMS sensors can be utilized in point-of-care testing (POCT) applications. POCT refers to diagnostic testing performed at or near the site of patient care, enabling rapid results and immediate clinical decision-making without the need for centralized laboratories. Such systems are increasingly important in applications ranging from infectious disease detection to chronic disease monitoring, especially in resource-limited or time-critical settings. MEMS technology offers unique advantages for developing POCT sensors, including miniaturization, high sensitivity, and potential for mass production at low cost, making it appealing for portable, fast, and accessible diagnostic platforms [14].

To detect the type and/or quantity of target molecules using MEMS-based biosensors, the chemical properties of bioparticles are initially converted into mechanical properties by the MEMS transducer, which is a movable part, such as a cantilever [15], [16], a fixed beam [17], or a membrane [18]. Depending on the output of the MEMS transducer, these sensors have two main operational modes: dynamic and static. In the dynamic mode, the mass of bioparticles induces a shift in the mechanical resonance frequency of the MEMS transducer. This frequency shift's magnitude determines the target molecules' type, quantity, or concentration [19]. The static mode-based biosensors utilize the mechanical deformation of the MEMS transducer caused by surface stress to measure the concentration of the target bioparticles [20], [21]. In this scenario, the target bioparticles in the sample are captured by bioreceptors immobilized on the surface of the mechanical transducer. This interaction causes a change in the surface stress, influenced by the binding of the bioreceptors to the target bioparticles [18], [22]. The resulting difference in surface stress between the top and bottom surfaces of the transducer leads to either tensile or compressive deformation, depending on the bond strength of the surface bioreceptors with the target molecules [23]. It is notable that surface stress-based sensors offer shorter response times and higher mechanical sensitivity compared to the other MEMS-based biosensors [24], [25].

In surface stress-based devices, various methods have been recently proposed for measuring the displacement of the movable part of the BioMEMS sensors, including capacitive [26], piezoresistive [27], and optical approaches [24]. Optical methods exhibit superior performance in terms of accuracy, sensitivity, detection limits, and resolution in comparison to other detection approaches.

In the optical detection methods, the position sensing detection method is one of the first introduced optical techniques to determine the deflection of the movable part of the surface stress-based biosensors [28]. To enhance the functionality of MEMS-based biosensors, new methods relying on optical interferometry, such as Fabry–Perot [29] and Mach–Zehnder [30], [31] interferometers, have been introduced. Furthermore, extensive efforts have been made to develop biosensors with optical detection methods based on

variations of the effective refractive index, such as surface plasmon resonance (SPR) [29], [30], optical phase shift [31], [32], [33], and metamaterial-based approaches [34]. These methods have led to significantly increased optical sensitivity. In these techniques, the absorption of target molecules by the biosensor results in a shift in the device's output transmission or reflection spectrum. However, the absorption of target molecules in these methods leads to very small changes in the effective refractive index. In these methods, BioMEMS sensors with phase shift detection can be extremely sensitive due to their ability to detect very small changes in the mode's refractive index [35]. This makes them suitable for detecting biomarkers at low concentrations [36]. Besides, they are less affected by noise compared to the other MEMS biosensor technologies, which is appealing to obtain more accurate results [37].

This article presents a surface stress-based BioMEMS platform that utilizes an optical detection system based on phase modulation to measure the concentration of the target biological particles. In this biosensor, the absorption of target bioparticles induces surface stress, leading to membrane deformation and altering the air gap between the waveguide and the movable component. As the membrane approaches the waveguide, there is a corresponding increase in the refractive index of the mode, which leads to a significant alteration in the phase of the mode. The variation in the phase shift is directly proportional to the displacement of the membrane and, consequently, the target bioparticles concentration in the test sample. The proposed biosensor offers ultrasensitive optical sensitivity, enabling the detection of specific analytes and accurately measuring their concentrations in samples for point-of-care (POC) tests.

The rest of this article is organized as follows. Section II presents the operational principles and analytical findings of the proposed device. The fabrication process is detailed in Section III. Furthermore, Section IV presents a comparative discussion of the operational characteristics of the current optical device with recent contributions in this area.

II. DESIGN AND SIMULATION OF THE PROPOSED BioMEMS SENSOR

A. Operational Principles

As shown in Fig. 1(a), the proposed device features a silicon membrane suspended above the silicon photonic waveguide that carries the guided mode with an initial gap of 300 nm. This membrane serves as the MEMS transducer, converting the absorption of target bioparticles into mechanical displacement. Both the membrane and waveguide are positioned on a SiO₂ layer. The membrane structure includes a precisely etched narrow strip at its center, directly above the waveguide. This etching is essential, as a thicker membrane could allow the propagating mode to leak into the membrane layer instead of remaining confined within the waveguide. By etching this section, the optical mode is effectively kept well-confined within the waveguide, preventing unintended propagation into the membrane. Table I outlines the geometrical and physical properties of the device.

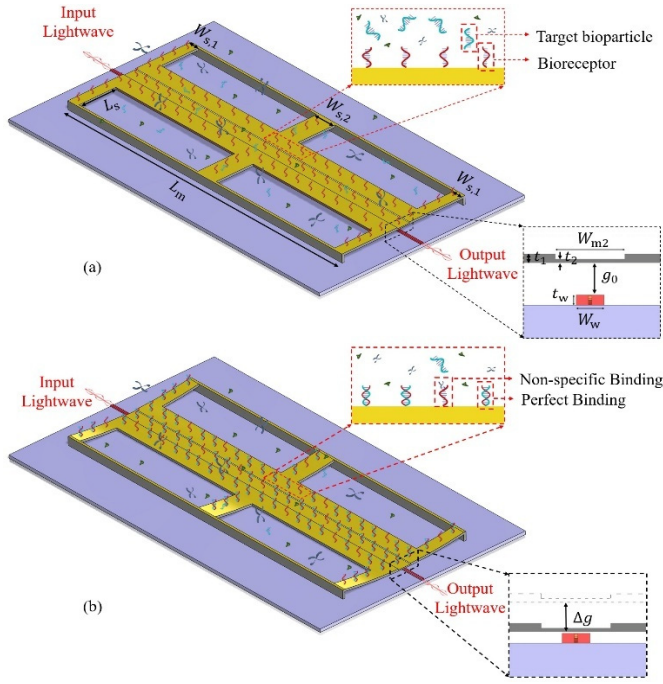


Fig. 1. Schematic of the proposed device, including (a) its physical parameters. (b) Deformation of the movable part after the interaction of target bioparticles with the immobilized bioreceptors.

TABLE I
DIMENSIONS AND PHYSICAL PARAMETERS

Symbol	Description	Value
E	Young's modulus of Si	165 GPa
ν	Poisson's ratio of Si	0.28
L_s	Length of stands	180 μm
m	Mass of the movable part	0.398 μg
k	Spring constant along the moving direction	0.464 N/m
t_1	Thickness of the stands and membrane	0.5 μm
t_2	Thickness of the etched portion	0.1 μm
W_{s1}/W_{s2}	Width of stand	45/90 μm
W_{m1}	Width of membrane	220 μm
W_{m2}	Width of the etching of the membrane	45 μm
L_m	Length of membrane	1500 μm
g_0	Initial gap	300 nm

To functionalize the sensing area, the membrane's surface is initially coated with a very thin layer of gold, approximately 10 nm in thickness, followed by the immobilization of bioreceptors on this surface. Each bioreceptor is supposed to capture a specific target molecule and, theoretically, does not interact with other molecules in the test sample. However, in practical applications, once the bioreceptors are immobilized, it is essential to cover the sensing area with blocking reagents such as bovine serum albumin, casein, or gelatin to prevent nonspecific binding on the sensing surface [38]. Consequently, as the test sample flows over the sensor's surface, the target particles are captured by the bioreceptors. The absorption of target molecules generates surface stress on the top surface of the sensor, leading to the displacement of the membrane's central strip, which can be calculated by

$$\Delta g = \left[\frac{3}{E(t_1)^2} + \frac{3W_{m2}}{W_{m1}E(t_2)^2} \right] (L_m + L_s)(1 - \nu)\Delta\sigma \quad (1)$$

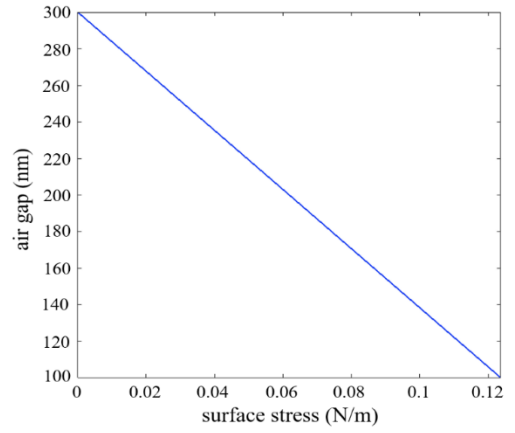


Fig. 2. Displacement of the moving membrane (Δg) versus the changes of the surface stress ($\Delta\sigma$).

where L_s , L_m , W_{m1} , W_{m2} , t_1 , and t_2 represent, respectively, the length of stands, length of membrane, width of membrane, width of the etched strip of the membrane, thickness of the membrane, and thickness of the etched strip of the membrane. Besides, E , ν , and $\Delta\sigma$ denote Young's modulus, Poisson's ratio of Si, and the induced surface stress [20].

In this design, the waveguide serves as a medium for confining and guiding the optical signal through the sensing region. Its geometry and material properties are optimized to enhance light-matter interaction, ensuring efficient sensing. As light propagates through the waveguide, a fraction of the optical mode extends beyond the core into the surrounding medium, forming the evanescent field. Any mechanical displacement of the membrane due to bioparticle absorption alters the boundary conditions of the evanescent field, leading to a phase shift in the guided mode as follows:

$$\Delta\phi = (2\pi/\lambda)L\Delta n \quad (2)$$

where L represents the phase shift length, Δn denotes the change in the refractive index of the mode, and λ refers to the optical mode wavelength. Ultimately, the phase shift observed at the output can be utilized to identify the concentration of the target bioparticles. The subsequent subsections discuss the operational features of the proposed sensor, derived from analytical and numerical analyses.

B. Mechanical Analysis

As previously noted, the absorption of target bioparticles in the sample by the bioreceptors immobilized on the top surface of the membrane induces surface stress, resulting in membrane deformation. This deformation alters the initial gap between the membrane and the waveguide carrying the propagating mode. The variations in the air gap, influenced by changes in surface stress, are depicted in Fig. 2. To achieve optimized performance, the dimensions of the proposed BioMEMS sensor have been designed through both analytical and numerical analyses, as summarized in Table I. By adjusting the surface stress within the range of 0–0.122 N/m, the air gap can be modified from 300 to 100 nm. Consequently, the mechanical

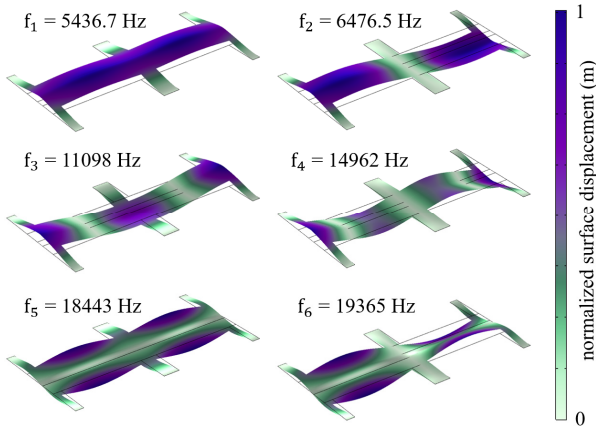


Fig. 3. Resonance frequencies of the proposed device, simulated by COMSOL Multiphysics.

sensitivity can be achieved as follows:

$$S_m = \frac{\Delta g}{\Delta \sigma} = 1.64 \mu\text{m/N} \cdot \text{m}^{-1}. \quad (3)$$

One of the important mechanical characteristics of MEMS sensors is the resonance frequency that can determine the device's bandwidth (BW). The frequency response of the MEMS sensor is obtained by analyzing its dynamic behavior using the equation of motion. Considering the sensor as a damped mass-spring system, the first resonance frequency can be determined as follows [39]:

$$f_r = \frac{1}{2\pi} \sqrt{\frac{k}{m}} \quad (4)$$

where k and m are the spring constant and the mass of the movable part of the MEMS biosensor, respectively.

Fig. 3 illustrates the first six mechanical resonance modes of this device, which was obtained through numerical methods, confirming the results from analytical approaches. The resonant modes of the proposed BioMEMS sensor were obtained through numerical simulations in COMSOL Multiphysics. The structure was modeled based on the elasticity equations and appropriate boundary conditions. By performing an eigenfrequency analysis, the natural frequencies and corresponding mode shapes were extracted. These modes are influenced by the sensor's geometry, material properties, and the induced surface stress due to bioparticle absorption. To determine the BW of the proposed sensor, the frequency response is also illustrated in Fig. 4. As shown in the figure, the frequency response curve remains completely flat over a broad range of frequencies before the first resonant mode. Therefore, the proposed biosensor BW can be determined as $\text{BW} = 2 \text{ kHz}$.

C. Optical Analysis

As previously noted, the proposed BioMEMS platform utilizes the phase modulation of the propagating mode to detect variations in the air gap. The interaction between the propagating mode and the movable membrane affects the effective refractive index due to changes in the evanescent field distribution. As the membrane displaces, the gap between the waveguide and membrane varies, altering the fraction

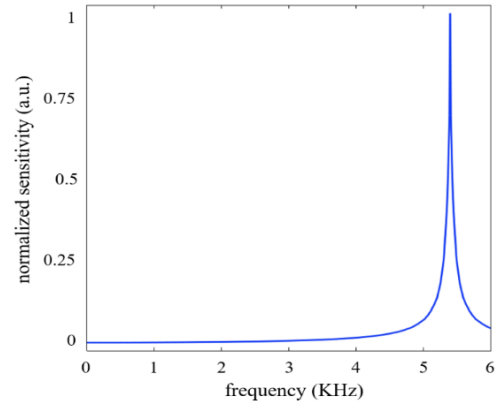


Fig. 4. Proposed device frequency response.

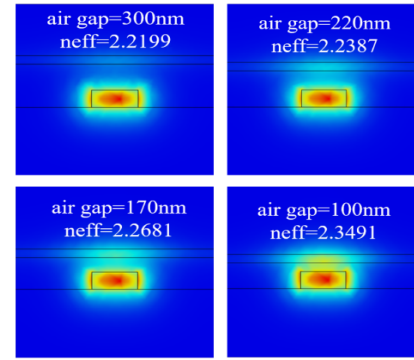


Fig. 5. Propagating mode's profile in different air gaps.

TABLE II
MATERIAL AND PHYSICAL CHARACTERISTICS OF THE
OPTICAL SENSING PART

Symbol	Description	Value
$n_{\text{eff,si}}$	Silicon refractive index	3.48
$n_{\text{eff,siO2}}$	Bottom cladding refractive index	1.44
L	Sensing length of the waveguide	1500 μm
W_w	Width of the waveguide	460 nm
t_w	Height of the waveguide	220 nm
λ	Operating wavelength	1550 nm

of optical power interacting with the surrounding medium (Fig. 5). This modifies the optical confinement conditions, leading to a phase shift in the propagating mode. To achieve an optimal balance between the optical sensitivity of the present design and its footprint, the dimensions of the optical sensing component have been optimized through finite element method (FEM) analysis, as detailed in Table II.

To assess the sensitivity of the proposed device, a numerical analysis is conducted to evaluate the phase shift of the propagating mode in response to changes in the effective refractive index. This analysis takes into account varying air gaps between the waveguide and the membrane (Fig. 6). As the air gap changes from 300 to 100 nm, the effective refractive index changes from 2.2199 to 2.3491. Thus, with a sensing length of 1500 μm at the operating wavelength of 1550 nm, a phase shift of 250π is obtained by (2). Therefore, the optical

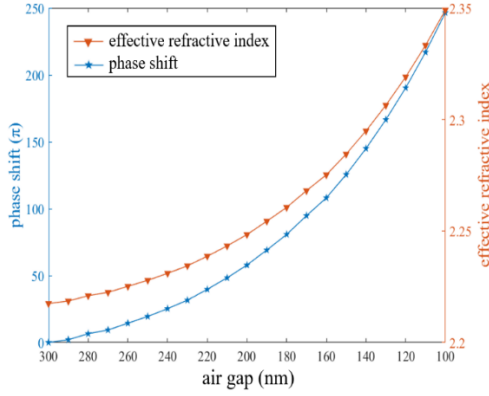


Fig. 6. Variation of the mode's phase and effective refractive index with the air gap change between the waveguide and membrane.

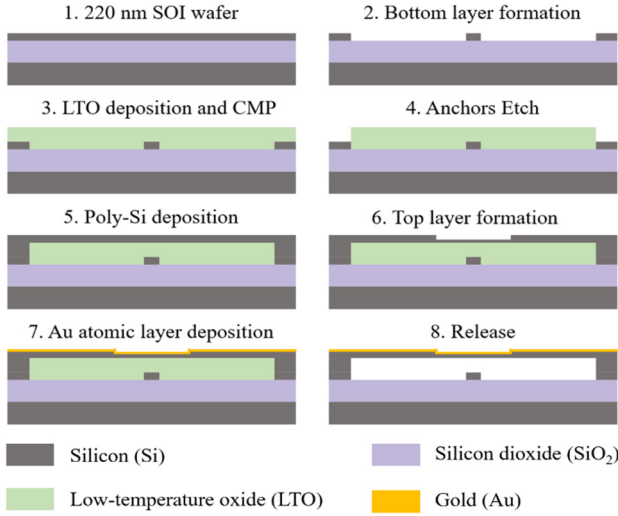


Fig. 7. Proposed fabrication process.

sensitivity of the present device, S_o , can be defined as follows:

$$S_o = \frac{\Delta\phi}{\Delta n_{\text{eff}}} = 1935 \pi \text{ rad/RIU}. \quad (5)$$

Finally, the length of the sensing region is also essential in designing BioMEMS sensors based on optical phase shift detection. To consider the length of the sensing region, one can introduce another parameter called the figure of merit (FOM) that can be calculated for the present BioMEMS device as

$$\text{FOM} = \frac{S_o}{L} = 1.29 \pi \text{ rad/RIU}\mu\text{m}. \quad (6)$$

III. PROPOSED PROCESS FLOW

The fabrication process, shown in Fig. 7, starts with a silicon-on-insulator (SOI) wafer comprising a 220 nm-thick silicon device layer, a 3 μm buried oxide (BOX) layer, and a silicon substrate. The optical waveguide and anchors are first defined by photolithography and fully etched through the 220 nm silicon layer using RIE etching. A 0.5 μm -thick low-temperature oxide (LTO, SiO_2) is then deposited by low-pressure chemical vapor deposition (LPCVD) at $\sim 400^\circ\text{C}$ – 450°C to act as a spacer layer. Chemical mechanical polishing (CMP) is applied to planarize the surface and reduce

TABLE III
FUNCTIONAL FEATURES OF THE PROPOSED DEVICE
COMPARED TO RELATED CONTRIBUTIONS

Functional characteristic	Proposed Bio-sensor	[41]	[35]	[40]
Optical sensitivity (rad/RIU)	1935π	1760π	1864π	316π
$\Delta\phi$ (rad)	250π	9.55π	10.6π	6.5π
L_{sensing} (μm)	1500	4100	7000	5000
Figure of merit (rad/RIU μm)	1.29π	0.429π	0.2662π	0.0632π

the LTO thickness to 0.3 μm . The anchor regions are opened by fully etching the LTO down to the silicon device layer.

An 800 nm amorphous silicon (a-Si) membrane layer is subsequently deposited via LPCVD. Its structural features, including the narrow strip and stands, are patterned using a two-step etching process: a 300 nm shallow etch followed by a 700 nm full etch. Finally, a 10 nm gold film is conformally deposited using atomic layer deposition (ALD) to enhance optical properties and, if needed, provide an electrode surface. This combination of lithography, etching, deposition, and planarization yields a suspended MEMS membrane precisely aligned with integrated photonic structures for optimal optical and mechanical performance.

IV. COMPARATIVE STUDY

In this section, several recent contributions [35], [40], [41] that rely on the same optical detection approach (i.e., phase modulation) have been compared to the proposed BioMEMS device. Several approaches [35], [40], [41] based on optical phase modulation suffer from small changes in the refractive index and, thus, low optical sensitivity while the target molecules are absorbed on their surface. The proposed micro-sensor, which operates on the principle of phase modulation, effectively addresses this challenge by delivering substantial optical sensitivity within a compact footprint [35], [40], [41]. Table III summarizes the comparative study between the proposed biosensor and several related contributions.

The data presented in this table demonstrate that the operational characteristics of the proposed device surpass those reported by previous studies. While the optical sensitivity of the device detailed in [35] and [41] is comparable to that of the present biosensor, the proposed BioMEMS sensor offers superior functional characteristics, particularly regarding sensing length and FOM. Ultimately, due to its remarkable attributes, including high sensitivity, rapid detection, compact and portable design, and label-free detection, the proposed sensor is appealing for POC testing applications.

V. CONCLUSION

In this article, we presented an ultrasensitive BioMEMS sensor that employs a MEMS transducer with an optical detection method based on phase shift modulation. The MEMS transducer features a silicon membrane coated with a thin layer of gold, onto which bioreceptors are immobilized to capture target biomolecules in the test sample. The absorption of these

target biomolecules alters the surface stress of the membrane, leading to its deformation and a change in the air gap distance between the movable component and the waveguide. As the air gap decreases, the refractive index of the guided mode increases, resulting in an optical phase shift that indicates the concentration of target bioparticles in the test sample. Our numerical and analytical results reveal the following functional features for the proposed BioMEMS device: a phase shift of 250π , an optical sensitivity of 1935π rad/RIU, a mechanical sensitivity of $1.64 \mu\text{m}/\text{N} \cdot \text{m}^{-1}$, and a FOM of 1.29π rad/RIU μm . These impressive functional characteristics suggest that the proposed BioMEMS platform can surpass existing technologies in terms of overall performance, making it particularly attractive for POC testing applications.

REFERENCES

- [1] M. Ahmadian and K. Jafari, "A graphene-based wide-band MEMS accelerometer sensor dependent on wavelength modulation," *IEEE Sensors J.*, vol. 19, no. 15, pp. 6226–6232, Aug. 2019.
- [2] E. Soltanian, K. Jafari, and K. Abedi, "A novel differential optical MEMS accelerometer based on intensity modulation, using an optical power splitter," *IEEE Sensors J.*, vol. 19, no. 24, pp. 12024–12030, Dec. 2019.
- [3] R. Gholamzadeh, K. Jafari, and M. Gharooni, "Design, simulation and fabrication of a MEMS accelerometer by using sequential and pulsed-mode DRIE processes," *J. Micromech. Microeng.*, vol. 27, no. 1, Jan. 2017, Art. no. 015022.
- [4] 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.
- [5] O. Legendre 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.
- [6] Y. Gholami, B. S. Darki, M. Moradi, K. Jafari, and M. H. Moaiyeri, "A multifunctional optical MEMS logic device for integration into reprogrammable photonic circuits for artificial intelligence applications," *Optik*, vol. 327, May 2025, Art. no. 172299.
- [7] K. Menon, R. A. Joy, N. Sood, and R. K. Mittal, "The applications of BioMEMS in diagnosis, cell biology, and therapy: A review," *Bio-NanoScience*, vol. 3, no. 4, pp. 356–366, Dec. 2013.
- [8] A. R. Kalaiarasi and G. P. Aishwarya, "Microsensor for cancer detection and MEMS actuator for cancer therapy," *Trans. Electr. Electron. Mater.*, vol. 24, no. 1, pp. 82–90, Feb. 2023.
- [9] Y. Wang, Y. Shi, and F. Narita, "Design and finite element simulation of metal-core piezoelectric fiber/epoxy matrix composites for virus detection," *Sens. Actuators A, Phys.*, vol. 327, Aug. 2021, Art. no. 112742.
- [10] Y.-J. Choi, T. Takahashi, M. Taki, K. Sawada, and K. Takahashi, "Label-free attomolar protein detection using a MEMS optical interferometric surface-stress immunosensor with a freestanding PMMA/parylene-C nanosheet," *Biosensors Bioelectron.*, vol. 172, Jan. 2021, Art. no. 112778.
- [11] N. Azizipour, R. Avazpour, D. H. Rosenzweig, M. Sawan, and A. Ajji, "Evolution of biochip technology: A review from lab-on-a-chip to organ-on-a-chip," *Micromachines*, vol. 11, no. 6, p. 599, Jun. 2020.
- [12] K. K. Shukla and T. Muthumanickam, "A smart sensor using MEMS technology for artificial environmental monitoring," *Mater. Today, Proc.*, vol. 66, pp. 3626–3633, Jan. 2022.
- [13] M. Jia, E. Zhongbo, F. Zhai, and X. Bing, "Rapid multi-residue detection methods for pesticides and veterinary drugs," *Molecules*, vol. 25, no. 16, p. 3590, Aug. 2020.
- [14] Y. Gholami, B. S. Darki, M. H. P. Isfahani, K. Jafari, and T. Azad, "An ultrasensitive MOEMS platform relied on an adjustable defect width for label-free bioparticle detection," *IEEE Sensors J.*, vol. 25, no. 10, pp. 17309–17316, May 2025.
- [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.
- [16] F. Marvi and K. Jafari, "A novel photonic crystal BioNEMS sensing platform based on fano resonances," *J. Lightw. Technol.*, vol. 39, no. 22, pp. 7296–7302, Nov. 1, 2021.
- [17] M. Khorsandifard, K. Jafari, and A. Sheikholeh, "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, Dec. 2021.
- [18] F. Marvi and K. Jafari, "A label-free biomarkers detection platform relied on a bilayer long-wave infrared metamaterials BioNEMS sensor," *Nanotechnology*, vol. 33, no. 26, Jun. 2022, Art. no. 265502.
- [19] S. Leahy and Y. Lai, "A gap method for increasing the sensitivity of cantilever biosensors," *J. Appl. Phys.*, vol. 122, no. 6, Aug. 2017, Art. no. 064502.
- [20] 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.
- [21] F. Marvi and K. Jafari, "A biosensing platform based on metamaterials BioNEMS for lab-on-chip systems," *IEEE Trans. Nanobiosci.*, vol. 23, no. 1, pp. 11–17, Jan. 2024.
- [22] F. Marvi and K. Jafari, "A novel BioNEMS sensor based on tunable plasmonic modes of a grating MIM structure," *IEEE J. Sel. Topics Quantum Electron.*, vol. 29, no. 4, pp. 1–8, Jul. 2023.
- [23] S. Sang, Y. Zhao, W. Zhang, P. Li, J. Hu, and G. Li, "Surface stress-based biosensors," *Biosensors Bioelectron.*, vol. 51, pp. 124–135, Jan. 2014.
- [24] 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.
- [25] R. Mathew and A. Ravi Sankar, "A review on surface stress-based miniaturized piezoresistive SU-8 polymeric cantilever sensors," *Nano-Micro Lett.*, vol. 10, no. 2, pp. 1–41, Apr. 2018.
- [26] V. Tsouti et al., "Evaluation of capacitive surface stress biosensors," *Microelectronic Eng.*, vol. 90, pp. 37–39, Feb. 2012.
- [27] 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.
- [28] 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.
- [29] X. Liu, M. Jiang, T. Dong, Q. Sui, and X. Geng, "Label-free immunosensor based on optical fiber Fabry–Pérot interferometer," *IEEE Sensors J.*, vol. 16, no. 20, pp. 7515–7520, Oct. 2016.
- [30] F. Vogelbacher et al., "Waveguide Mach–Zehnder biosensor with laser diode pumped integrated single-mode silicon nitride organic hybrid solid-state laser," *Biosensors Bioelectron.*, vol. 197, Feb. 2022, Art. no. 113816.
- [31] M. J. Goodwin, G. A. J. Besselink, F. Falke, A. S. Everhardt, J. J. L. M. Cornelissen, and J. Huskens, "Highly sensitive protein detection by asymmetric Mach–Zehnder interferometry for biosensing applications," *ACS Appl. Bio Mater.*, vol. 3, no. 7, pp. 4566–4572, Jul. 2020.
- [32] S. Lin et al., "Using antibody modified terahertz metamaterial biosensor to detect concentration of carcinoembryonic antigen," *IEEE J. Sel. Topics Quantum Electron.*, vol. 27, no. 4, pp. 1–7, Jul. 2021.
- [33] V. Yesudasu and H. S. Pradhan, "Performance enhancement of a novel surface plasmon resonance biosensor using thallium bromide," *IEEE Trans. Nanobiosci.*, vol. 21, no. 2, pp. 206–215, Apr. 2022.
- [34] F. Marvi, K. Jafari, and S. Aziminam, "A dual-axis metamaterial BioNEMS accelerometer for biomedical applications," *IEEE Sensors J.*, vol. 23, no. 7, pp. 7041–7048, Apr. 2023.
- [35] Q. Liu et al., "Highly sensitive Mach–Zehnder interferometer biosensor based on silicon nitride slot waveguide," *Sens. Actuators B, Chem.*, vol. 188, pp. 681–688, Nov. 2013.
- [36] P. K. Sahoo, S. Sarkar, and J. Joseph, "High sensitivity guided-mode-resonance optical sensor employing phase detection," *Sci. Rep.*, vol. 7, no. 1, p. 7607, Aug. 2017.
- [37] J. Hong et al., "A Mach–Zehnder interferometer based on silicon oxides for biosensor applications," *Analytica Chim. Acta*, vols. 573–574, pp. 97–103, Jul. 2006.
- [38] M. Salehizadeh, P. Dehghani, M. Zimmermann, V. A. L. Roy, and H. Heidari, "Graphene field effect transistor biosensors based on aptamer for amyloid- β detection," *IEEE Sensors J.*, vol. 20, no. 21, pp. 12488–12494, Nov. 2020.
- [39] W. Zhang et al., "Structural optimization of the micro-membrane for a novel surface stress-based capacitive biosensor," *Microelectronic Eng.*, vol. 106, pp. 9–12, Jun. 2013.

- [40] Y. Liang, M. Zhao, Z. Wu, and G. Morthier, "Bimodal waveguide interferometer RI sensor fabricated on low-cost polymer platform," *IEEE Photon. J.*, vol. 11, no. 2, pp. 1–8, Apr. 2019.
- [41] L. Laplatine et al., "Silicon photonic olfactory sensor based on an array of 64 biofunctionalized Mach–Zehnder interferometers," *Opt. Exp.*, vol. 30, no. 19, pp. 33955–33968, 2022.



Yashar Gholami received the master's degree in electrical engineering from Shahid Beheshti University (SBU), Tehran, Iran, in 2019, where he is currently pursuing the Ph.D. degree.

His research interests include the design and modeling of microelectromechanical sensors and actuators, photonic integrated circuits, and nanophotonic devices.



Zahra Alinia received the B.S. degree in electrical engineering from the Firouzabad Institute of Higher Education, Firouzabad, Iran, in 2019, and the master's degree in electrical engineering from Shahid Beheshti University, Tehran, Iran, in 2024.

Her research interests include MOEMS sensors, optical sensors, and photonic integrated circuits.



Behnam Saghirzadeh Darki received the Ph.D. degree in electrical engineering from Isfahan University of Technology (IUT), Isfahan, Iran, in 2017.

Since 2017, he has been an Assistant Professor at the Department of Electrical Engineering, Islamic Azad University, Najafabad, Iran, where he is currently the Manager of economic, investment, and knowledge-based affairs.



Kian Jafari (Senior Member, IEEE) joined the Faculty of Engineering and the Interdisciplinary Institute for Technological Innovation (3IT), Université de Sherbrooke, Sherbrooke, QC, Canada, as an Associate Professor, in 2023. He has a solid background in micro/nanosystems (M/NEMS) and strong publication records in highly ranked international journals. Over the last years, he developed several micro/nanodevices for various applications such as biomedical instruments and AI systems.

Mr. Jafari was various prestigious grants, including the NSERC Discovery Grant and the Canadian Cancer Society Innovation Grant, to support his research endeavors.



Mohammad Hossein Moaiyeri (Senior Member, IEEE) is an Associate Professor with the Faculty of Electrical Engineering, Shahid Beheshti University, Tehran, Iran. He has authored or co-authored more than 100 research articles in international journals. His current research interests include very large-scale integration (VLSI) design for beyond-CMOS emerging devices, low-power VLSI design, and mixed-signal integrated circuits.

Dr. Moaiyeri has also been an Associate Editor of *IEEE ACCESS* and *Circuits, Systems, and Signal Processing* journals.