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Bouras Mounir
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Abdesselam Hocini

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Analysis of highly sensitive biosensor for glucose based on a one-dimensional photonic crystal nanocavity

Bouras Mounir,* Charik Haouari, Allal Saïd, and Abdesselam Hocini
Mohamed Boudiaf University of M'sila, Department of Electronics, Signal and Systems Analysis Laboratory, M'sila, Algeria

Abstract. A one-dimensional (1-D) photonic crystal (PhC) composed of multilayer thin films having the form $(Si/Air)_9$ is investigated. The selection of Si and air is because their use as a material platform provides the ability to fabricate sensors with other photonic devices on a single chip. We design a biosensor concept that uses an optical resonance in a 1-D silicon PhC with an air defect, which is theoretically studied for refractive index (RI) sensing in the infrared wavelength region (700 to 1900 nm). A similar crystal with a defect layer by changing the dielectric constant in different orders is also studied. A method is presented to maximize the sensitivity of the biosensor and to obtain a very narrow bandwidth cavity mode for good biosensor resolution. We optimized the thickness of the air defect layer in the PhCs on both sides of the nanocavity and show that a high sensitivity value of 1118 nm/refractive index unit (RIU) with a detection limit of 0.001 RIU, which proves the ability of the design to produce biosensor PhC. This optical biosensor has a very much higher sensitivity and simpler structure compared to other PhC biosensors reported previously. [DOI: 10.1117/1.OE.58.2.027102]

Keywords: optics one-dimensional photonic crystals; biosensor; microcavity; waveguide; refractive index.

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1 Introduction

The photonic crystals (PhCs) are periodicities of dielectrics that can control the propagation of the electromagnetic wave traveling through them.[1] The periodic variation of their physical properties (dielectric constant) leads to photonic bands of wavelengths at which the light can propagate or reflected.[2,3] PhCs have been used in a wide range of applications, such as bend waveguides, filters, sensors, lasers, amplifiers, and resonators.

Diverse structures have been investigated as optical sensors, such as fiber-assisted surface plasmon resonances,[4] fiber Bragg gratings,[5] step-index fibers and PhC fibers,[6] fiber-assisted surface plasmon resonances,[4] and PhCs.[7–9] Optical resonance in PhCs has known an increasing development for refractive index (RI) biosensing applications in recent years.

The PhCs-based structures with one-dimensional (1-D) and two-dimensional (2-D) bandgaps are very promising for microfluidics biosensing as the defect layer can be injected with analytes that can provide variation in RI and hence manipulate the dispersion of PhC waveguides.[10] Various designs based on a 2-D PhC slab with a triangular arrangement of holes in an silicon-on-insulator substrate have been used for optofluidic sensing with a sensitivity of 636 nm refractive index unit $(RIU)^{-1}$.[1] A new design L7, the defect line is modified and optimized, which leads to a sensitivity $(\Delta\lambda/\Delta n)$ of 510 nm RIU^{-1}, has been achieved compared to a sensitivity of about 100 nm RIU^{-1} for L3 cavities of water and ethanol solutions.[9]

Biosensors based on 2-D PhC optical sensors incorporating with microcavities have many advantages in compactness, high sensitivity, and quality factor (QF). But the fabrication of these sensors is difficult and complicated, and the sensitivities that can be achieved by these optical sensors are not high. Because of easy fabrication, 1-D PhCs have attracted a lot of attention for fabrication of different components in silicon-based photonic integrated circuits at IR wavelength.[8]

One-dimensional PhCs, or dielectric multilayer structures, have been used in a wide range of applications, such as filters,[11] microcavities,[12] and optical sensors.[12,13] The configuration based on the excitation of surface electromagnetic waves in a 1-D PhC biosensor has been proposed,[11–13] which proves the sensitivity is much higher compared to the sensitivity of conventional surface plasmon resonance biosensors.[8]

In this paper, a 1-D dielectric multilayer nanocavity is investigated for RI sensing of glucose solution at an IR wavelength of 1.55 μm. In present structure, Si and air layers are used to achieve high contrast between the RI of the high index and low index adjacent layers and to have a strong optical field in the free space region of the air nanocavity. In addition, we used different values of defect width D.

The simulation results show that a change in ambient RI is apparent; the sensitivity of the sensor is achieved. We show that a high sensitivity of >1118 nm RIU^{-1} with a detection limit of 0.001 RIU can be obtained for RI biosensing of glucose solution in the IR. The calculations were carried out using rigorous coupled-wave analysis (RCWA) and enhanced with modal transmission-line (MTL) theory, which is a rigorous, fully vectorial solution of Maxwell's equations (RSoft CAD).

2 Theory and Principle

The RCWA method uses the concept of a unit cell to handle both 2-D and three-dimensional periodic structures and is specifically tailored for multilayer structures. The unit cell

*Address all correspondence to Bouras Mounir, E-mail: mouno_25000@yahoo.fr

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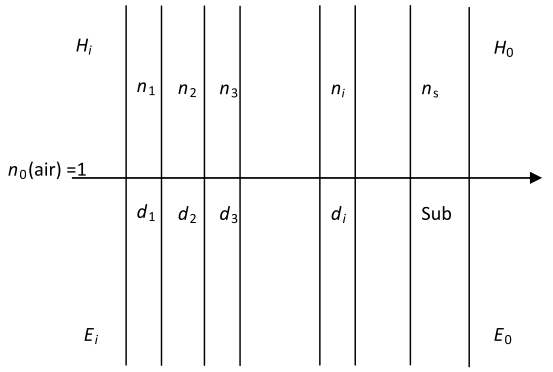

Fig. 1 System of multilayer film (1-D PhC).

definition can have arbitrary geometry, and the index distribution can consist of both standard dielectric materials and dispersive or lossy materials, such as metals. The input or incident plane wave can have arbitrary direction and polarization.

RCWA[14–16] represents the electromagnetic fields as a sum over coupled waves. A periodic permittivity function is represented using Fourier harmonics. Each coupled wave is related to a Fourier harmonic, allowing the full vectorial Maxwell's equations to be solved in the Fourier domain.

The MTL[17] approach is equivalent to RCWA, except that MTL uses summations over individual modes to represent the electromagnetic fields. This method has the advantage of describing the fields in terms of an equivalent transmission-line network, which yields physical insight for various applications using different mechanisms. Another benefit of the MTL method is that it eliminates possible stability problems, which may be encountered in other numerical schemes, such as the transfer matrix method.

Maxwell's equations can be expressed as

$$\frac{\partial H_x}{\partial t} = -\frac{1}{\mu}\frac{\partial E_z}{\partial y}, \tag{1}$$

$$\frac{\partial H_y}{\partial t} = \frac{1}{\mu}\frac{\partial E_z}{\partial x}, \tag{2}$$

$$\frac{\partial E_z}{\partial t} = -\frac{1}{\varepsilon}\left(\frac{\partial H_y}{\partial x} - \frac{\partial H_x}{\partial y} - \sigma E_z\right), \tag{3}$$

where ε is the permittivity, μ represents the permeability, and σ is the conductivity.

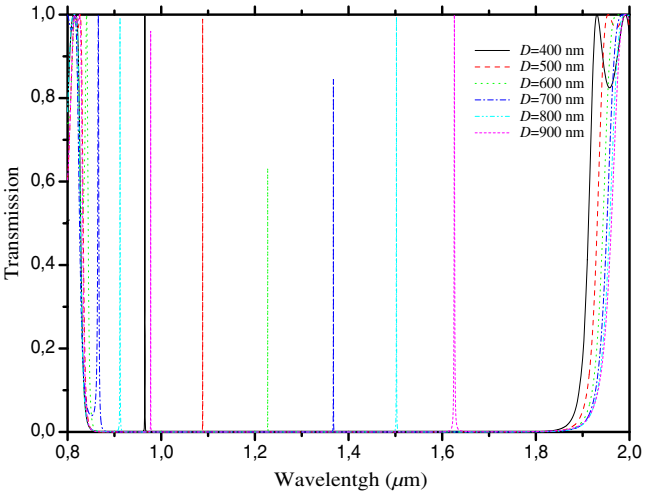

Fig. 3 Transmission of the 1-D PhC air nanocavity for $d_{Si} = 0.10$ μm and $d_{air} = 0.20$ μm. Different modes oscillate for central air defect width (D) of 400 to 900 nm (from left to right).

In this simulation, the structure is actuated with a Gaussian pulse with transverse magnetic (TM) mode and transverse electric (TE) mode into the first waveguide, and then, the resonant wavelength is detected from the end of the second waveguide. The light is incident perpendicularly from the air ($n_o = 1$) and propagates throughout the layers of the multilayer thin film, then it is finally transmitted in a glass substrate ($n_s = 1.52$), then to air again (shown in Fig. 3).

The spatial and temporal steps are related through the following equation:

$$\Delta t \leq \frac{1}{C}\sqrt{\Delta x^{-2} + \Delta y^{-2}}, \tag{4}$$

where Δx and Δy are spatial steps in the x and y directions, respectively, and c is the speed of light in vacuum.

The utilization of microfluidic channels provides spatial confinement of the analyte directly on the reaction site. Bonding to a borosilicate glass top wafer proved inefficient to prevent fluidic leakages into the adjacent waveguides regions, thus making it more difficult to pump the fluid in the microchannel.[18] This issue was addressed by bonding a polydimethylsiloxane (PDMS) lid on top of the structured wafer. PDMS is able to fully seal the fluidic channel and prevent leaks along the waveguides, due to its elastomeric properties. In this way, planarization by chemical mechanical

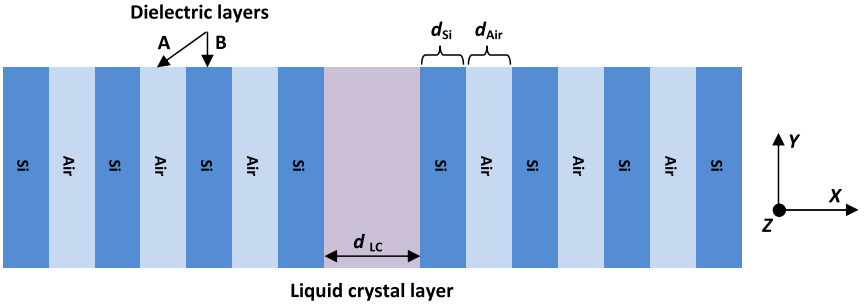

Fig. 2 One-dimensional PhC nanocavity.

polishing was avoided, which would significantly complicate the fabrication process.

3 Results and Discussion

The considered array of layers takes the form (AB)9, shown in Fig. 1, where A denotes Si and B denotes air. Silicon and air layers are used to achieve high contrast of the RI between the high index and the low index adjacent layers. Figure 2 shows the suggested 1-D PhC biosensor is constructed by making a central defect of width D in the middle air gap filled with the analyte (glucose solution) in water of RI[18]

$$n = 1.33313 + 0.001 * C, \tag{5}$$

where C is the concentration of the glucose in percent. The sensitivity of the suggested 1-D PhC biosensor is studied in both TE and TM modes. To study this structure for glucose biosensing applications, the thickness of the silicon wall d_{Si} and the width of the air gap d_{air} are fixed to

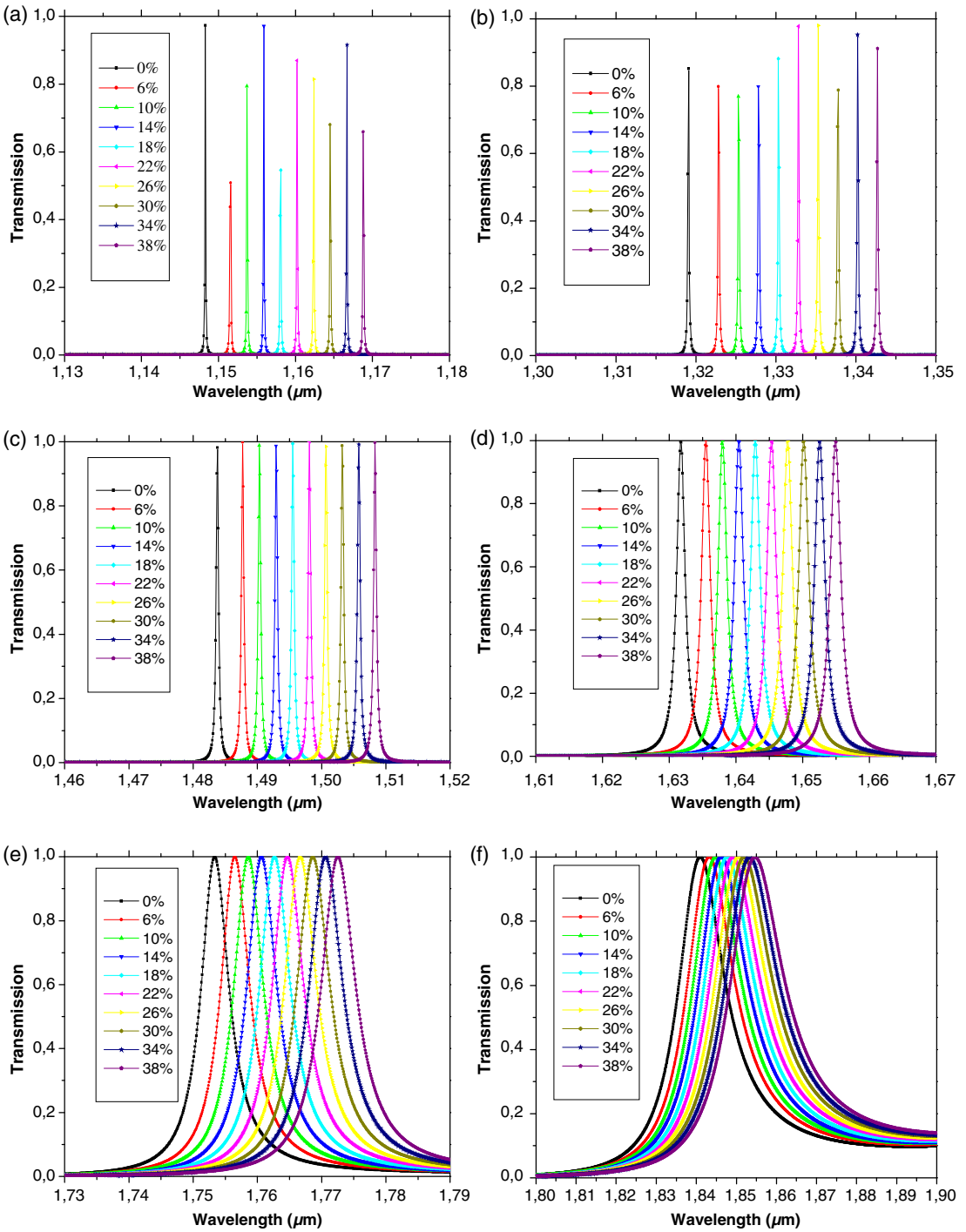

Fig. 4 TE transmission spectra of the proposed 1-D PhC biosensor with defect widths: (a) $D = 500$ nm, (b) $D = 600$ nm, (c) $D = 700$ nm, (d) $D = 800$ nm, (e) $D = 900$ nm, and (f) $D = 1000$ nm.

the values 100 and 200 nm, respectively. When we choose the thickness of the Si layer $d_{Si} = 100$ nm, the thickness of the air layer $d_{air} = 200$ nm, and the cavity length $d_{cavity} = 700$ nm, only one cavity mode is excited at a wavelength of 1.55 μm.

The sensitivity of the PhC biosensor is defined as the resonance wavelength (allowed modes) shift $\Delta\lambda_{max}$ per the change in the analyte RI Δn (RIU)

First, the 1-D PhC air nanocavity is simulated using the RCWA method. The method is used to calculate the reflection and transmission properties of the cavity and gives the amplitude of waves reflected by and transmitted through the cavity.

When we choose the thickness of the Si layer $d_{Si} = 100$ nm, the thickness of the air layer $d_{air} = 200$ nm, and the central air defect width (D) is changed from 500 up to 900 nm with a step of 100 nm, only one cavity mode is excited at a wavelength of 1.55 μm. The transmission of the cavity is shown in Fig. 3, for different cavity widths (D). By changing the cavity width, the wavelength of the cavity mode can be adjusted in the large bandgap a (shown in Fig. 3) from 0.7 to 1.9 μm. As the central defect width D changes, the transmission of the traveling electromagnetic wave through the defected region will be changed.[19]

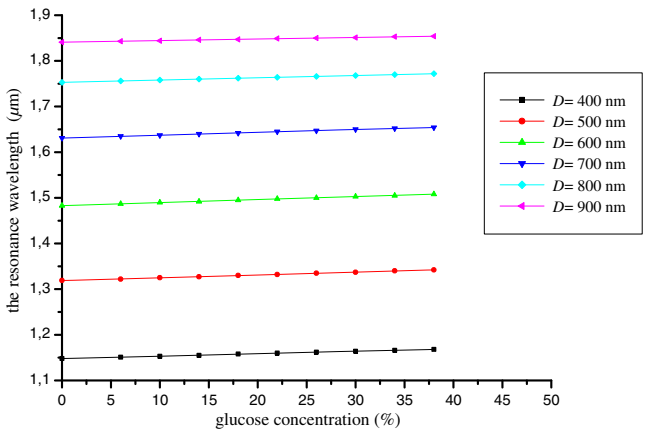

Fig. 5 The shift in the resonance wavelength as a function of the urea concentration at different values of defect width D in case of TE mode.

Table 1 The change in sensitivity of the suggested 1-D PhC biosensor with different values of the defect's width D.

	Sensibility (nm/RIU)	
Defect's width D (nm)	TE	TM
400	936	810
500	1068	867
600	1118	919
700	1054	911
800	868	827
900	622	598

Figures 4 and 5 show TE transmission spectra through the proposed 1-D PhC for different values of defect width D starting from 400 up to 900 nm with a step of 100 nm at different glucose concentrations. It is revealed from these figures that the change of the defect's width due to changes in the analyte RI is good.

In the case of TE mode, the best sensitivities of 1068, 1118, and 1054 nm/RIU are at both defect's width values $D = 500$ nm, $D = 600$ nm, and $D = 700$ nm, respectively as shown in Figs. 4(b)–4(d).

Further, at the values of $D = 400$ nm, the TE transmission spectra show lower maximum transmissions at resonance wavelengths for some values of glucose concentration as shown in Fig. 4(a). Moreover, as shown in Figs. 4(e) and 4(f), the maximum transmission at resonance wavelengths is closing to the unity at $D = 800$ and 900 nm.

Therefore, very high sensitivity can be obtained for the suggested design. Table 1 shows the variation of the resonance wavelength with the glucose concentration at different values of defect width D. It is evident from this figure that the resonance wavelength is slightly affected by the change in the glucose concentrations.

In fact, the sensitivity of the PhC biosensor is not only the most important factor in the efficiency of the 1-D PhC biosensor. The degree of defect modes sharpness and then the QF are required to test the resolution of the suggested 1-D PhC biosensor. The QF of the suggested PhC biosensor is defined as

$$\mathrm{QF} = \lambda_0/\Delta\lambda_{\mathrm{FWHM}}, \tag{6}$$

where λ_0 is the resonance wavelength and $\Delta\lambda_{\mathrm{FWHM}}$ is the full width at the half maximum of the output transmission. The cavity wavelength and QF have been plotted in Fig. 6, as a function of the cavity length. The QF has values of 5800 and 5200 at defect widths $D = 500$ and 600 nm, respectively, and as it has values of 21 and 32 at defect widths $D = 800$ and 9000 nm in TE mode.

The sharpness of the peaks increases with changing the change of the defect's width because the number of interfering beams increases inside the 1-D PhC biosensor. Additionally, at some values of urea concentrations, the increment of defect's width leads to minimize the maximum

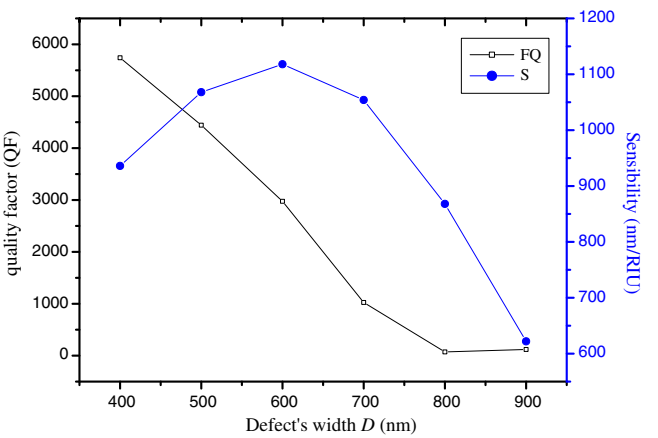

Fig. 6 Effect of the number of (Si/air) periods N on the sensitivity (black line) and the QF (blue line).

Table 2 Comparison of the proposed sensor with various PhC designs.

References	Type of device	Sensibility (nm/RIU)	Δn
20	RI biosensor formed by two waveguide and microcavity	330	0.001
9	Sensor RI based of L3 PhC cavity	460	0.038
7	PhC cavity biosensor	510	0.001
1	PhC cavity biosensor	636	0.001
22	RI biosensor formed by two waveguide and microcavity	530	0.001
23	PhC 1-D cavity biosensor (*S. Uree*)	712	0.008
8	PhC 1-D cavity biosensor (C. gaz)	1200	0.001
This work	PhC 1-D cavity biosensor (S. glucose)	1118	0.001

transmission values at the corresponding resonance wavelength. The offered QF at $D = 600$ nm is equal to 3000. Moreover, the effect of the change of the defect's width on the sensitivity of the proposed 1-D PhC is very weak as shown in Fig. 6 (black line). There is a slight decrease in the sensitivity from 1200 nm/RIU at $D = 600$ nm to 620 nm/RIU at $D = 900$ nm. In comparison with other biosensor using PhCs,[20,21] our designed sensor has higher sensitivity (Table 2).

4 Conclusion

A 1-D dielectric multilayer nanocavity is investigated for RI sensing of glucose solution at an IR wavelength of 1.55 μm. An air defect is used to create a cavity in the PhC.

The principle of detection is based on the resonance shift of the cavity as a function of the RI, which leads to a shift in the output transmission spectrum.

Due to the high contrast between the refractive indices of Si and air central defect of 1-D PhC, defect modes appear, and the position of which depends on the width of air defecting layer (D) also increases the sensitivity of the sensor.

A method was presented to maximize the sensitivity and resolution of the sensor. By optimizing the central air defect width in the 1-D PhCs, we are able to obtain a high sensitivity of 1118 nm RIU^{-1} and a narrow bandwidth cavity mode for high biosensor resolution. The biosensor optimized here has a simple structure and can be integrated into silicon-based photonic circuits.

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Bouras Mounir received his MSc degree in electronics from the Department of Electronics, University of Science and Technology Houari Boumediene, Algeria, in 2010 and his PhD in electronics from the University of M'sila in 2016. Currently, he is a member of the Laboratoire d'Analyse des Signaux et Systemes. In 2001, he joined the University of M'sila, Algeria, as an assistant lecturer.

His scientific interests are optical waveguides and optical communications.

Charik Haouari obtained the master's degree in electronics from the Department of Electronics, University of M'sila, in 2018 in instrumentation.

Allal Saïd obtained the master's degree in electronics from the Department of Electronics, University of M'sila, in 2018 in instrumentation.

Abdesselam Hocini received his MSc degree in electronics from the Department of Electronics, University of Constantine, in 2003 and his PhD in electronics in 2008. In 2001, he joined the University of M'sila, Algeria, as an assistant lecturer. In 2016, he obtained a title of professor. Currently, he is the head of the Laboratory Laboratoire d'Analyse des Signaux et Systemes. His current research interests are modeling and simulation and study of optic propagation in materials.