On the Enhancement Anomaly Detection for RF Bio-Sensors by Computing Artificial Networks Using Machine Learning Techniques

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***Abstract*—This work proposes serval sensor designs for a low-cost, highly-sensitive microwave sensor for identifying different liquid samples by monitoring the variation in S21 magnitude. The sensor is developed using an interdigital capacitor (IDC) in series connection with a circular spiral inductor (CSI) and connected directly to a photo-resistor (LDR). To enhance sensor insertion losses, the sensor is introduced to a Hilbert fractal open stub and coupled to an interdigital capacitor to operate at 1.22GHz. The accuracy of the sensor is significantly improved using a back loop trace, eliminating nonlinear effects from multi-layer diffractions. An analytical model based on circuit theory is suggested for the proposed sensor operation. The authors found an observable influence of varying the LDR value on sensor insertion losses, motivating the development of the sensor prototype. The sensor is manufactured and tested experimentally before and after samples introduction, with a human glucose sample mounted on the LDR patch to measure the effects of light intensity.**

***Index Terms*—Microwave sensor, nondestructive biomedical measurements, Hilbert fractal structure, circular spiral, glucose, neural network.**

I. Introduction

Microwave sensors play a crucial role in biomedical applications, including noninversion sensing processes and analysis of bio tissue dielectric properties [1]. The future outlook for medicine is directed towards personal treatment regimens, which aim to establish personalized treatment plans for each patient [2]. Microwave technologies offer low-cost and low-power sensitivity, especially in the complex combination of compounds found in human body fluids like blood, glucose, and spinal fluids [3]. This leads to a wide time gap between sample acquisition and associated results. The increasing prevalence of chronic diseases and the need for cost-effective healthcare are the main challenges facing researchers in this field [4]. To achieve good medical care at low costs, researchers are focusing on spreading awareness of prevention and effective treatment against diseases rather than focusing on advanced treatment systems [5]. Healthcare providers and employers are increasingly adopting modern communication technologies, such as microwave sensors, to promote "e-health" monitoring, which provides accurate and early diagnosis without the need for external medical control [6].

Biosensors for microwaves depend on the characteristics that determine the electromagnetic fields interacting with materials based on their molecular structure [7]. Microwave biosensors are designed to mutate changes in wave spread speed during the biological environment into a quantifiable signal, providing the diversity of a specified bio-parameter [8].

RF/microwave resonators are essential in radio frequency (RF)/microwave frameworks for detection and quantization. For optimal channel plow insertion loss (IL), high return loss (RL), high frequency selectivity, low losses, more fetched, and compactness [6]. Microwave sensors advanced work with new sensing technologies to do multi-band operations. Such technology will make it possible for short-range, high-information-rate connections [8]. Microchannel planning commonly uses microwave resonators based on microstrip lines because of their cost-effectiveness, simplicity, and ease of manufacture [5]. To make multi-band microwave resonators, designers have used several different techniques, including stepping-impedance microstrip resonators, multi-mode resonators, parallel-coupled line resonators, and transmission zeroes [9]. To make sensors simpler and smaller, we can use stepped-impedance resonators, which work well with two or more transmission lines that have different characteristic impedances [10].

Biosensors are essential components in medical and biological experiments and diagnostics, measuring the dose of various biochemical species in aqueous solutions [11]. They have become essential in fields such as diagnostics, pharmaceutical procedures, biomedical engineering, industry, agricultural, and food safety [12]. Researchers have proposed various techniques and results for detection in different materials. In [13], a waveguide cavity-based sensor was presented for measuring the concentration of liquid solutions. The sensor operates at 1.91GHz in the fundamental TE101 resonant mode and has been tested on water-sodium chloride and water sucrose combinations. However, the accuracy of the resonant frequency evaluation is directly related to the quality factor QF and conditions around the measured records. In [14], a substrate integrated wave-guide (SIW) sensor was proposed to measure the permittivity of liquids. The resonant parameters of the sensor are affected by liquids passing through a slot

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opened on the upper side of the sensor. The sensitivity sensor is tested on different percentages of water in ethanol. The arterial neural network was used to solve the problem with relative errors of 5% and 7%, respectively. In [15], a microwave resonator based on Complementary Circular Spiral Resonator (CCSR) was proposed, working at 4.72GHz. Through experimentation and analysis, the proposed sensor can determine the concentrations of ethanol-water mixtures by measuring the resonant frequency of the CCSR and the permittivity of sample under test (SUT). In [16], a microwave resonator based on the complementary split-ring resonator (CSRR) coupling with a microstrip and the microfluidic channel was proposed, working at 3.994GHz. The sensor can also determine the water content of glucose by measuring the resonant frequency of the CSRR and the permittivity of SUT. In [17], a microwave sensor with a coplanar waveguide semi-lumped meandering open complementary split ring resonator (MOCSRR) was proposed, operating up to frequencies of approximately 200MHz. The sensor successfully detects branded and unbranded fuel oil samples, with the difference demonstrated by the fluctuation in the transmission coefficient resonant frequency amplitude. In [18], a Cesare fractal geometry based on a compact Electromagnetic Bandgap (EBG) structure was presented to measure the complex permittivity of various liquids (butan-1-ol, methanol, and water). The relative permittivity is 3.57 for butan-1-ol, 21.3 for methanol, and 78 for water.

In this work, the proposed sensor is realized for enhancing anomaly detection of RF bio-sensors using machine learning techniques. For this the theoretical considerations and design are discussed in section II. The experimental validation is considered in section III. In section IV, the neural network implementation is realized. The paper is concluded in section V.

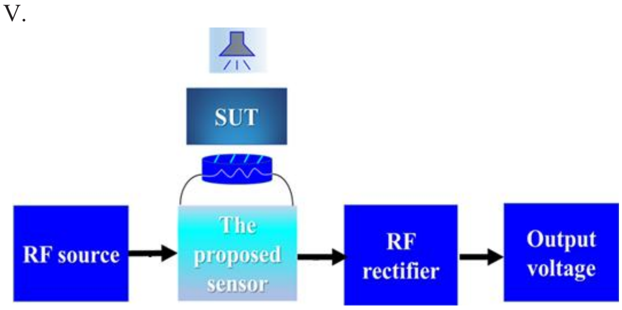


Fig.1: The simplified block diagram of the design based on reconfigurable technology.

II. THEORETICAL STUDY TO DESIGN A MICROWAVE SENSOR FOR BIOMEDICAL DETECTIONS

This paper presents new microwave sensors for biomedical detection using a two-port network for liquid characterizations. The main structure is a miniaturized microwave resonator based on a circular spiral inductor, which is used to increase sensitivity and concentrate current before being transferred to the SUT. The sensor is introduced to three inclusions: open stub, Minkowski filter, and Hilbert filter. Open stubs are used to eliminate measurement distortions and dispersion effects, while Minkowski filter is used to linearize measurements and remove the effects of frequencies before 0.5GHz. The Hilbert curve is used to remove the effects of frequencies before 0.5GHz. The sensors are designed to be compatible with microprocessors and measure water content variation based on frequency shift. The research also explores the use of a photo resistor to control sensor performance and measure output voltage from a RF rectifier at the sensor's output port. A neural network model is presented to solve problems involving nonlinearities, multi-variables, and multi-resonances.

A. Base Sensor Design

The proposed sensor is mounted on a FR4 substrate with a thickness of 1.6 mm and is based on OS-CRLH. It consists of a transmission line connected to an RLC branch network, which is structured as input capacitor which are inspired from [19] in series with a circular spiral inductor Lse and connected directly to a photo-resistor LDR. The LDR is mounted between the LC branches at the middle of the sensor, and the back panel is covered with a metallic ground plane of 0.035 mm. The proposed sensor geometrical details are explained as seen in Fig.2. Consequently, the main structure of the proposed sensor is constructed from the same proposed sensor with Hilbert curve introduction. As maintained later, the proposed sensor is constructed from an inter digital capacitor to remove the effects of the imaginary component that is generated by the indictor structure which mainly stores the energy from the propagation [19]. The inductor structure is invoked to be a spiral geometry cause of being a highly sensitive stricter that concentrate the current before being transferred to SUT.

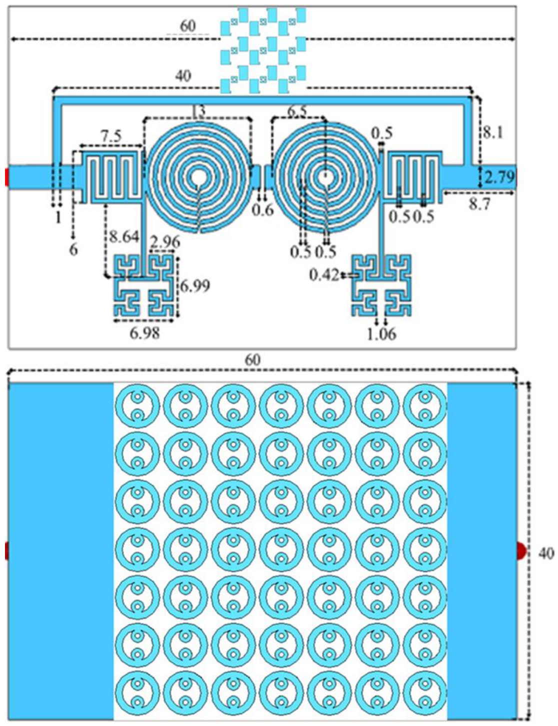
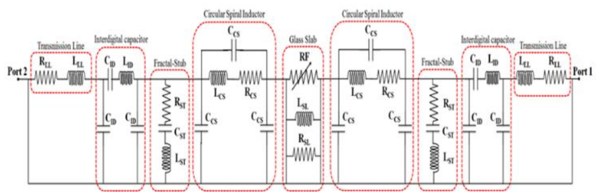


Fig.2; The proposed sensor geometry: (a) front view and (b) back view. Note: all dimensions are in mm scale.

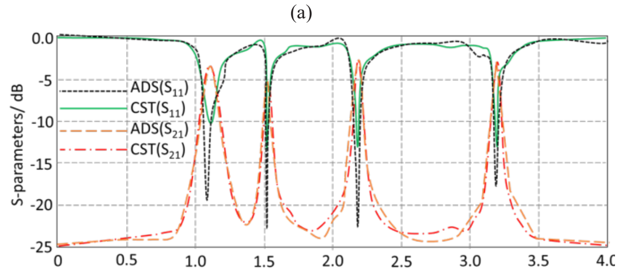
The suggested design examines the Hilbert-curve configuration to minimize sensor dimensions and enhance frequency resonances [20]. The examined fractal geometry is founded on the third-order Hilbert geometry, as seen in Fig. 2.

The suggested equivalent circuit model for Hilbert geometry, as shown in Fig. 2, is developed from Richard Koch's theory [8]. The equivalent circuit model of the proposed unit cell is produced from the simulation results, as seen in Fig. 2. The equivalent capacitances of coupling between the unit cell and neighboring cells are denoted as the left-hand capacitor (CST), while the fractal slot is regarded as an inductor (LST), where the magnetic current mobility in the air traces may be amplified. This inductor corresponds to the magnetic field contained inside the fractal slots of the rings. The material loss is determined by the resistor RST [11]. The suggested sensor design has a feedback loop structure. The benefit of this introduction is to get a band reject filter response rather than a passband resonance. This alteration enhances the accuracy of the measurement findings [12].

The suggested sensor design is founded on an equivalent circuit model, which is analytically derived from an analogous circuit model based on the established RLC network, typically configured as an IDC in series with a CSI and the Hilbert fractal. The suggested structural equivalent circuit model is developed with the lumped elements Richard model [1]. The suggested circuit model consists of a 50Ω input impedance RF source connected in series with a parallel (R-L-C) branch, as seen in Fig.3. The primary transmission line was characterized by an inductive segment LT and capacitive air gaps Cgap, as previously seen in Fig.3(a). The proposed circuit model's S-parameters, shown in Fig. 3(b), are analyzed and juxtaposed with those obtained via CST MWS. An effective agreement is attained based on the specified lumped components, which are modeled in Advanced Devices Simulator (ADS). The assessed RLC components are enumerated in Table I.



(a)



(b)

Fig.3 Equivalent circuit analytical model of the proposed sensor: (a) circuit model and (b) S-parameters.

TABLE I
LUMPED ELEMENT VALUES OF THE EQUIVALENT CIRCUIT MODEL

Eleme nt	RL H	RR H	GL H	GR H	CL H	CR H	LL H	LRH
Valus	12.2 Ω	50 Ω	0.1 S	4 S	1.1p F	3.1p F	3nH	2.2n H

C. Sensor Operation and Detection Process

This section presents the operational and reconfiguration scenarios designed to illustrate the proposed technique for comprehending sensor functionality. By altering the resistance of the included LDR from 100Ω to 1000Ω, a notable shift in the spectra of the suggested sensor S21 is seen, as shown in Fig. 4. This change is ascribed to the voltage division effects between the overall impedance of the proposed sensor and the LDR [13]. The present motion would be substantially influenced and traverse the back loop structure to be diffused inside the suggested fractal form. Consequently, this dissipation will be very beneficial for sensing, as will be shown subsequently. Fig.4 illustrates three frequency resonances at 630MHz, 1.22GHz, and 3.2GHz, all of which will be used in the sensing procedure of this study.

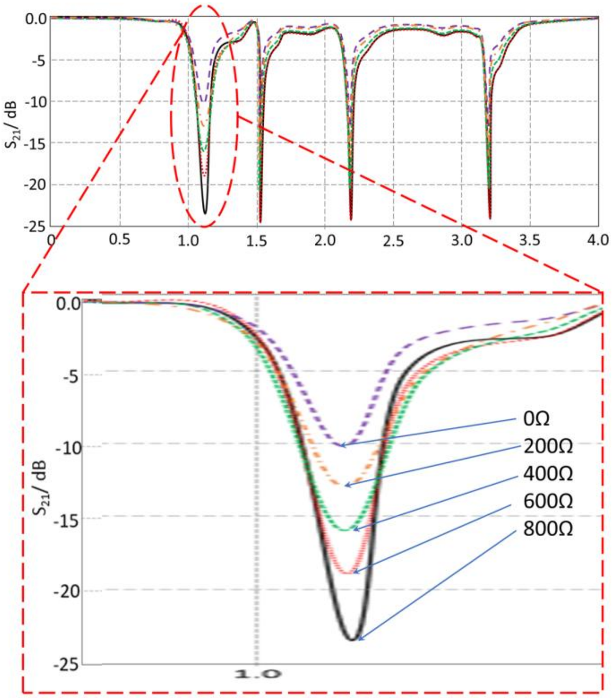


Fig.4: Evaluated S_{21} spectra with respect to varying the LDR value.

III. EXPERIMENTAL VALIDATIONS

This paper presents a sensor design for glucose samples aimed at improving detection accuracy by generating an output voltage. The main limitation is the difficulty in linearizing measurements due to field fringing from boundary conductions. The proposed method uses a photo-resistor to control sensor performance, eliminating field fringing effects from discontinuities. The sensor is connected with a back loop structure, allowing for band reject filter response and controlling charging rise time until the photo-resistor reaches saturation. The sensor is fabricated and measured experimentally using a Professional Network Analyzer (Agilent PNA 8720) after a through transmission calibration process. The measurements are conducted to S11 and S21 spectra to eliminate possible errors. The sensor shows a well-defined resonance at 1.5GHz with S21= -27 dB. An excellent agreement is found between numerical results and measurements within frequencies from 0.1GHz up to 4GHz.

The sensor was tested by taking different samples of the glucose and water mixture and measured experimentally in terms of S21 spectra. The use of an LDR switch is conducted with a long glass slab to avoid direct contamination and linearize variation using the LDR switch. The sensor is fabricated using a wit chemical etching process and S21 spectra are measured before and after glucose introduction on top of the LDR spot using an Agilent vector network analyzer. For this, an experimental study was conducted to validate the effectiveness of glucose level variation on sensor performance at 1.22GHz. The results showed that the proposed sensor is an excellent candidate for glucose measurements and could be promising for other biological fluid characterizations. The sensor was introduced to 15 patients and measured glucose levels at three different times, about 7 days to 10 days apart. The results showed that the variation in output voltage increased rapidly with increasing glucose magnitude as listed in Table II. The glucose sample used was about 0.01cc to avoid contamination effects.

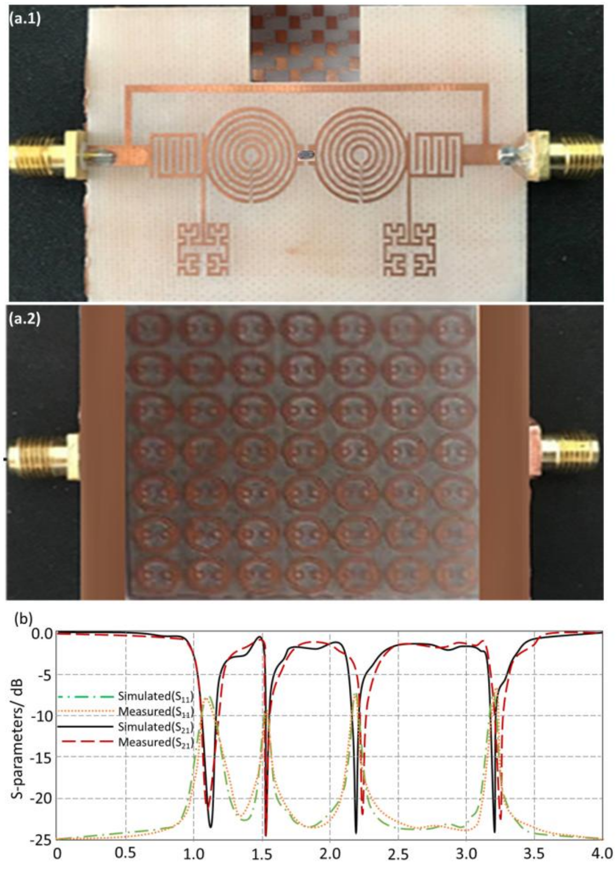


Fig.5; Experimental validation: (a) Fabricated sensor and (b) S-parameter spectra.

The proposed sensor is linked to an RF rectifier to gauge the output voltage across a resistor connected in parallel to the output terminals of the RF rectifier, as shown in Fig. 6(a). This measurement is conducted by varying the light intensity on the photoresist from 100 Ω to 600 Ω. This variance is analyzed quantitatively by examining the effective impedance change and its impact on the S21 value at 1.22GHz. The fluctuation is experimentally monitored about an output voltage (Vout) for practical use. The fluctuation in Vout has a linear trend, as seen in Table II. This signifies that a satisfactory concordance has been attained between the experimental and simulated outcomes. The proposed sensor variation in response may be attained by using an RF rectifier to assess the alteration in output voltage via an oscilloscope. The input voltage is set at 100 mV from the sources. Fig.6(b) illustrates the fluctuation of the output voltage of the proposed RF rectifier in relation to the change in input power.

TABLE II
MEASURED GLUCOSE INFLUENCE ON THE PROPOSED SENSOR PERFORMANCE.

Case number	BMI	Age/ year	Sex	Glucose level	mV	|S21|
1	19.1	7	F	102	72.2	0.106
				200	74.7	0.115
				156	73.9	0.111
2	24.9	51	F	111	72.8	0.109
				123	72.3	0.112
				201	75.3	0.115
3	21.4	70	M	300	77.2	0.121
				125	72.9	0.111
				245	76.3	0.156
4	28.1	45	M	301	76.9	0.116
				359	78.1	0.182
				277	78.4	0.174
5	26.7	56	M	231	74.5	0.123
				93	72.1	0.193
				108	72.4	0.191
6	32.1	35	F	122	73.1	0.103
				133	73.1	0.109
				143	73.9	0.108
7	33.5	43	F	185	74.1	0.122
				164	73.9	0.133
				166	72.4	0.124
8	29.8	42	F	101	73.1	0.091
				91	72.9	0.094
				98	73	0.092
9	23.4	49	F	144	74.1	0.091
				187	77.8	0.098
				145	77.4	0.092
10	32.6	37	M	190	76.3	0.109
				123	75.6	0.106
				144	76.1	0.11
11	36.1	72	M	102	72.4	0.111
				300	77.3	0.11
				340	77.9	0.113
12	34.6	67	M	390	77.8	0.189
				331	77.3	0.188
				301	77	0.177
13	21.4	63	M	210	74.9	0.109
				243	75.2	0.101
				226	75.1	0.105
14	22.5	68	M	189	74.2	0.195
				130	73.4	0.196
				210	74.4	0.179
15	23.8	54	M	221	75.3	0.109
				289	75.9	0.11
				234	75.2	0.112

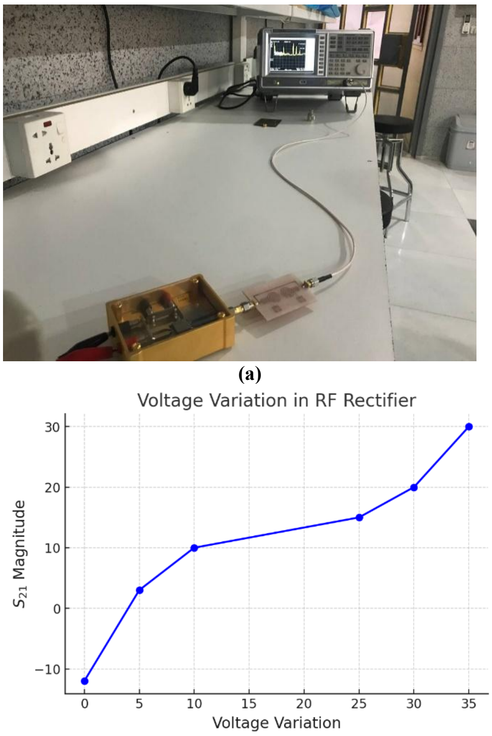


Fig.6: (a) Experimental setup and (b) RF rectifier performance.

IV. NEURAL NETWORK IMPLEMENTATION

In this section, the variation in the measured S_{21} magnitude from is performed. The S_{21} variation with respect to the glucose level normalization are evaluated from measurement data. In such data, the detection process is performed according to the S_{21} change. For this, a comparison study between measured data is conducted to realize the trade-line regressions for measurements with a negative slop as seen in Fig.7.

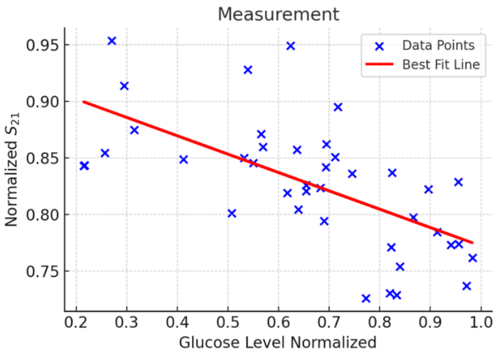


Fig.7: A comparison study between measured points.

This study used the instances from Table II to derive the input data. The samples are categorized into training and testing sets. To find the best performance index (P.I.) for the ANN, the number of neurons, epochs, and learning rate will be changed. The neural network will then be trained five times, and the results will be averaged. There is one buried layer containing three neurons. The learning rate is set at 0.001, with a total of 88 epochs. The mean P.I. of this network is 76.32%. The MATLAB code is used to categorize the input data from regression, demonstrating the categorization of the data based on their respective categories. The input data in this category is classified by the neural network into three periods, mostly based on the regression rate. In this categorization, the first third of the input data is designated as low glucose level. The second interval pertains to the intermediate glucose concentration. The last interval is regarded for elevated glucose levels. Figure 8 demonstrates that the data regression aligns very well with the output data. Furthermore, the regression topic is notably relevant; the suggested sensor, based on the neural network, achieves exceptional alignment with classifications at both low and high glucose levels. This finding is confined to intermediate values, which may result in significant inaccuracy. To provide an effective solution, more data points are necessary to identify the optimal fit for this period. It is noteworthy that the disparity in the intermediate period is shown in Fig. 7, which depicts a breakpoint in the center of the values from the simulation that aligns with the actual data. Table III enumerates the optimal values achieved for the most frequently used neural network parameters.

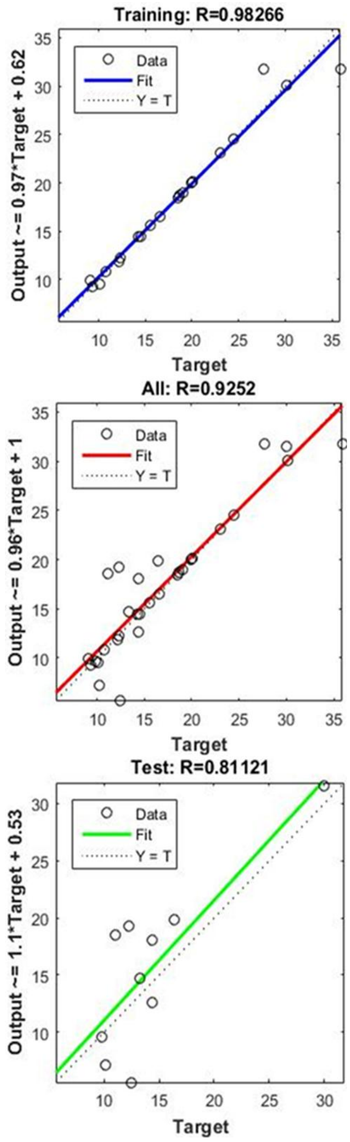


Fig.8: Findings from the regression analysis.

TABLE III
NN parameters

NN parameters	NN for simulation
Number of Input layer nodes	5
Number of Hidden layer nodes	3
Number of Output Layer nodes	1
Transfer function	logsig
Training function	pureline
Learning rate	0.001
Maximum number of Epochs	88

Finally, the proposed sensor is compared to the relatives published in the literature as listed in Table IV. The proposed work is compared interms of Q-factor, sensing type, and frequency. It is found that the proposed sensor provides the highest Q-factor with a resonance frequency of 1.22GHz. This frequency band is highly suitable for biomedical applications, particularly glucose level detection. Because it can balance penetration depth, sensitivity, and signal integrity, the 1.22GHz frequency is perfect for biomedical applications that need to find glucose levels [12]. Its dielectric properties, which vary with frequency, make it easier to detect variations in glucose concentrations. The frequency also keeps signal loss to a minimum, which makes it possible to use a non-invasive method that is still sensitive to changes in glucose levels. Additionally, it offers biomedical safety and non-invasiveness, as it doesn't require direct contact with blood. The high-Q sensor at 1.22GHz allows for compact microwave sensors with strong resonance characteristics, improving measurement precision.

TABLE IV
A comparison between the proposed sensor and other published results

Ref.	Q-factor	Sensing	Fo/GHz
[4]	280	Solid	2.4
[5]	407.34	Solid	2.2
[6]	345	Solid	3.2
[7]	652	Solid	2.22
[8]	446, 506	Solid	2.5 and 3.9
[9]	458	Solid	1.5
[10]	662	Solid	2.4
[11]	265	Liquid	2.45
[12]	398	Liquid	1.8, 2.45, and 3.5
[13]	425	Liquid	5.3 and 5.8
[14]	280, 160	Liquid	5.76 and 7.85
[15]	111.56, 21.39	Liquid	2.45 and 5.8
[16]	286.5	Powder	1 to 3
[17]	385.6	Powder	1.0–3
The proposed work	794.7	Liquid	1.22

V. Conclusion

The proposed sensor design utilizes a microwave resonator based on the CRLH structure of Hilbert geometry. It is tested with glucose from 11 different patients, and the way it works involves a new way of using an LDR part that makes the sensor work differently when the amount of glucose changes. We attribute this change to the transparency of the glucose under test, which alters the frequency shift and S21 magnitude. The writers found that the ratio in the S21 magnitude is critical at 1.22GHz with linear variation. Because of the linear variation, the authors believe that this design is ideal for sensing. The proposed sensor circuit model is used to see what happens when the proposed sensor parts are added, and the outcomes are contrasted with the actual outcomes that were measured. We numerically test the sensor's performance using CST MWS and validate it with ADS. It was solved analytically with circuit model analysis, and the suggested sensor works in a straight line, but the way it works changes depending on what is being tested. Transmission line technology forms the suggested sensor. It has a transmission line that is linked to an RLC network, set up as an IDC in series with a CSI, and linked directly to an LDR. The sensor operates at 1.22GHz and detects water introductions successfully by changing the S21 magnitude directly.

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