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Comparative Study of Jeans and FR4 Patch Antennas for Noninvasive Blood Glucose Sensing

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ABSTRACT: This paper presents a comparison between jeans and FR4-based patch antenna sensors for blood glucose sensing using a noninvasive approach. The dual-feed square-patch antenna sensor with partial ground has a compact structure operating at 2.8 GHz frequency. The performance of the antenna sensor was evaluated by measuring the shifts in the resonant frequency with varying blood glucose concentrations in the simulations. Both antenna sensors were experimentally tested and validated by placing a human volunteer's fingertip on the patch. The jeans-based antenna sensor exhibited higher sensitivity to glucose variations than the FR4 based antenna. This study demonstrated the crucial role of substrates in sensing applications involving interactions with human tissues.

1. INTRODUCTION

Heart rate, blood pressure, blood glucose, etc. are some of the basic human vitals that should be monitored regularly to ensure a healthy life. Diabetes Mellitus (DM) is a long-term metabolic disorder caused by an increase in blood glucose levels due to insulin deficiency or resistance [1]. Uncontrolled diabetes can lead to complications, such as cardiovascular disorders, nerve damage, kidney dysfunction, and vision impairment; therefore, the regular monitoring of blood glucose levels is crucial [2]. Different types of antennas, such as microstrip patch antennas, slot antennas, split ring resonators, and dielectric resonator antennas are used for blood glucose monitoring applications. These antenna-based sensors offer advantages such as compact size, low cost, high sensitivity, and ease of integration into wearable systems, thereby making them suitable for continuous glucose monitoring [3].

Different methods to measure blood glucose levels are categorized as invasive, minimally invasive, and noninvasive. In traditional invasive methods, fingers are pricked to take blood samples, and devices, such as Continuous Glucose Monitors (CGMs), involve skin penetration to measure glucose levels in the blood or interstitial fluid [4]. These methods are painful, increase the risk of infection, and inconvenient, leading to reduced patient compliance. Owing to these limitations, ongoing research is focused on noninvasive glucose measurement techniques that offer pain-free, continuous, and real-time glucose tracking [5, 6]. Noninvasive techniques include optical, electrochemical, and electromagnetic (EM) sensing. EM sensing offers good sensitivity owing to its better penetration into human tissue and less scattering in free space. The rise or fall in blood glucose alters the permittivity and conductivity of blood owing to variations in free water molecules in the blood. Thus, the propagation characteristics of radio frequency (RF) waves are affected by shifts in resonance frequency, reflection coefficient, and transmission characteristics, which can be measured using RF-based sensors such as patch antennas [7].

Recently, many studies have reported noninvasive glucose sensing using microwave and antenna-based sensors. A compact FR4-based narrow-band patch antenna for blood glucose monitoring was presented in [8]. The antenna operates at 6.1 GHz with a gain of 3.3 dBi and dimensions of $30 \times 30\,\text{mm}^2$. A human Finger Phantom Model (FPM) was utilized to test glucose sensitivity, and a maximum frequency shift of 800 MHz was observed when the glucose level was changed from 0 to 500 mg/dL. Further experimental validation using a glucometer is required. In [9], the authors presented an arrow-shaped patch antenna sensor on an FR4 substrate at 2.4 GHz, with compact dimensions of $35 \times 13.5 \times 1.6\,\text{mm}^3$. The resonant frequency showed measurable shifts when the FPM was placed 0.5 mm above the patch and 1.5 mm away from the feeding port. Although a sensitivity of 17 MHz/mg/mL was achieved for glucose concentration variations from 0 to 500 mg/dL, precise finger placement poses challenges for repeated measurement. Furthermore, in [10], the authors presented a planar Yagi-Uda antenna sensor operating at 5.5 GHz with a gain of 6.74 dBi and compact dimensions of $30 \times 40 \times 1.6\,\text{mm}^3$. The maximum frequency deviation obtained was 26 MHz for glucose concentrations between 250 mg/dL and 500 mg/dL. In [11], the authors attempted to improve the sensitivity of an antenna for glucose monitoring using an Electromagnetic Band Gap (EBG) structure and Complementary Split-Ring Resonator (CSRR) structures in the antenna design operating over 24.1 GHz to 29.3 GHz frequency. Even though a high sensitivity of 50 MHz/ε_r was achieved, the inclusion of an EBG structure increases the design complexity. In [12], a compact antenna sensor with dimensions of $20 \times 30\,\text{mm}^2$ was constructed using an FR4 substrate at 5 GHz with a quality factor of 471.

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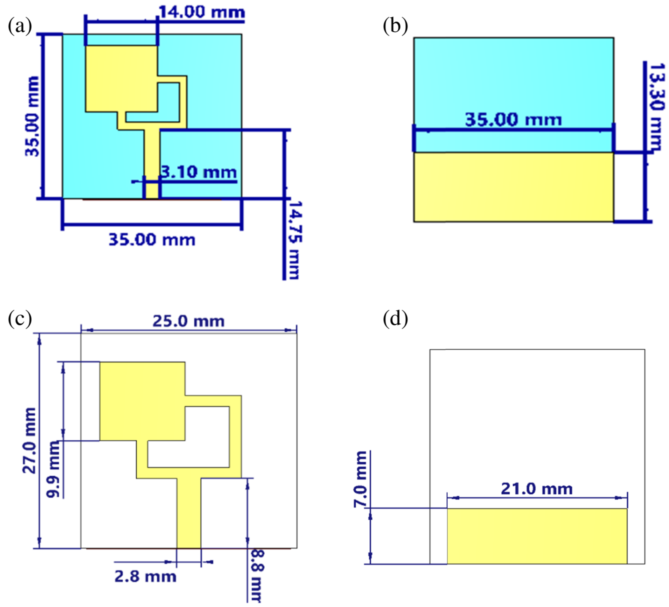

FIGURE 1. Geometry of the proposed antenna: (a) top view and (b) bottom view using jeans substrate, (c) top view and (d) bottom view using FR4 substrate.

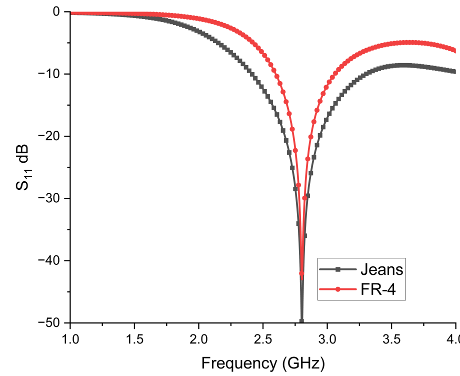

FIGURE 2. S_{11} versus frequency response of the proposed antennas.

Glucose measurements from 0 to 1000 mg/dL were performed using a finger phantom, and the total shift obtained was 24 MHz with a sensitivity of 0.48×10^{-3} (mg/dL). The blood sugar values were compared with glucometer readings, and a measurement error of approximately 1.9875% was obtained.

The authors in [13] designed a patch antenna on a Jeans substrate at 1.3 GHz to monitor high, normal, and low glucose levels. The antenna was simulated in the High Frequency Structure Simulator (HFSS) software on both the finger and arm phantom models. The maximum frequency shift achieved was 9 MHz for the arm phantom, whereas for the finger phantom, the frequency shift was 47 MHz. Here, the practical verification of the simulated results is missing. In [14], the authors presented a monopole-embroidered textile antenna operating at 2.4 GHz. The functionality of the antenna was tested by using a blood-mimicking aqueous solution, and the sensitivity was 350 kHz/(mg/dL). In [15], a compact Defective Ground Structure (DGS)-based glucose sensor was proposed for a frequency range of 1.9–2.3 GHz. The glucose concentrations considered in the experiment ranged from 0 mg/dL to 4800 mg/dL, and the frequency sensitivity was 1.852×10^{-5} GHz/(mg/dL). As the human blood glucose level remains below 500 mg/dL, the sensor exhibited low sensitivity at lower glucose concentrations.

Thus, there is a need for a high-precision and reliable antenna sensor to detect frequency shifts with varying glucose concentrations. The measurement accuracy of antenna-based sensors for noninvasive blood glucose monitoring applications depends on tissue properties (skin thickness, water content, etc.), environmental factors (temperature, humidity, noise, etc.), and sensor design (geometry, substrate properties, operating frequency, etc.), which are analyzed through parameters such as S-parameters, resonant frequency shifts, and changes in phase/amplitude. Most of the research has focused on simulation data for glucose sensing, and real-time measurements have not been implemented. For noninvasive blood glucose monitoring antenna sensors, the choice of substrates is crucial, as the accuracy of the sensor output depends on the substrate, along with the design of the antenna. The novelty of the proposed work lies in the application-oriented comparison between jeans (flexible) and FR4 (rigid) substrates, illustrating their effect on sensing behavior and thereby providing guidelines for the choice of substrates for wearable sensing applications.

2. ANTENNA DESIGN

The proposed antenna structure consists of a dual-feed square patch with a partial ground plane to achieve compactness. The partial ground plane is incorporated into the antenna geometry to achieve miniaturization, improve bandwidth, and enhance fringing fields. Other techniques, such as shorting pins and slot loading, also help in realizing a compact structure but lower gain and bandwidth, whereas stacked patches, metamaterial structures, and meandering techniques increase the design complexity, surface current losses, fabrication challenges, etc. [16]. The patch antenna was optimized and designed on two different substrates, namely jeans and FR4, operating at 2.8 GHz using the Computer Simulation Technology (CST) Studio Suite software. Jeans substrate, with a loss tangent of 0.025, thickness of 1.25 mm, and dielectric constant of 1.9, results in a larger structure ($35 \times 35\,\text{mm}^2$) than FR4 ($27 \times 25\,\text{mm}^2$), which has a higher dielectric constant of 4.3 and a loss tangent of 0.025. With an increase in the dielectric constant, the patch size decreases [17]. Fig. 1 shows the top and bottom topologies of the proposed antenna using jeans and FR4 as substrates. The idea of dual feeds is to create dual polarization and avoid finger misalignment issues for the proposed application of glucose monitoring.

The square patch was fed with two orthogonal $100\,\Omega$ feedlines that were added into $50\,\Omega$ feedline. The feeds were designed using the following equations [18]:

$$B = \frac{60\pi^2}{Zo\sqrt{\varepsilon_{eff}}} \tag{1}$$

$$W_z = \frac{2h}{\pi}\left\{B - 1 - \ln(2B-1) + \frac{\varepsilon_{r-1}}{2\varepsilon_r}\left[\ln(B-1) + 0.39 - \frac{0.61}{\varepsilon_r}\right]\right\} \tag{2}$$

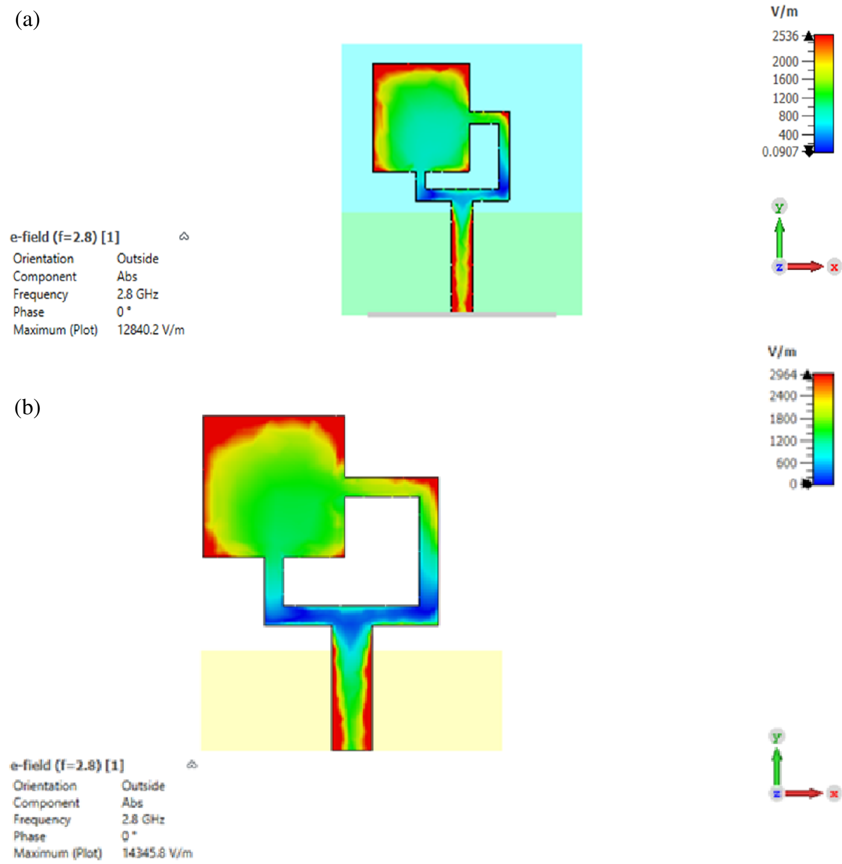

FIGURE 3. E-field distribution of the proposed antenna for (a) jeans substrate, (b) FR4 substrate.

where ε_{eff} represents the effective dielectric constant, ε_r the dielectric constant of the substrate, h the thickness of dielectric substrate, and Zo the impedance of the transmission line.

The proposed antennas resonate at 2.8 GHz, but the antenna designed on a jeans substrate exhibits better impedance matching and wider impedance bandwidth than FR4 substrate as depicted in Fig. 2. Ideally, the low dielectric constant and porous structure of the Jean fabric enhance the fringing fields around the patch edges, leading to a broader bandwidth. The Electric (E) field distribution for the two antennas is illustrated in Fig. 3, which confirms that the patch radiates efficiently with a strong E-field distributed over the edges of the patch and the feedline. The field strength along the y-axis and x-axis on the patch surface is strong enough for the interaction with the fingertip, facilitating noninvasive glucose sensing.

3. RESULTS AND DISCUSSION

In this section, the antenna performance is evaluated using a finger phantom model and human volunteers. The term finger phantom model refers to a multi-layered tissue-equivalent model of a human finger designed in simulation software to study antenna sensor-tissue interactions, rather than a medical measurement technique. For validation, the simulated results were compared with the fabricated antenna ones.

3.1. Antenna Performance on Finger Phantom Model

To evaluate the antenna performance for noninvasive blood glucose monitoring, an FPM was designed in CST software. The rectangular FPM is composed of different tissue layers, that is skin, fat, blood, and bone layers, to mimic a real human finger. The height of the finger was 12 mm, with a length of 30 mm, as illustrated in Fig. 4.

The dielectric properties and dimensions of the layers are listed in Table 1. This is a widely used FPM for the proposed application. The blood dielectric values changed with changes in the glucose concentration (Table 2). Therefore, these values were modified according to the change in glucose level from 0 to 500 mg/dL. The frequency-dependent dielectric properties of biological tissues are described by the mathematical model Cole-Cole [19] and are widely used in the literature. The equation is given as

$$\varepsilon^*(\omega) = \varepsilon_\infty + \frac{\varepsilon_s - \varepsilon_\infty}{1 + (j\omega\tau)^{1-\alpha}} + \frac{\sigma}{j\omega\varepsilon_0} \tag{3}$$

TABLE 1. Dielectric properties of the finger model [9].

Parameters	Skin	Fat	Blood	Bone
Dielectric constant	38	6	ε_r	11.5
Loss tangent	0.32	0.18	0.38	0.35
Dimension (mm^2)	12×12	10×10	9×9	4×4

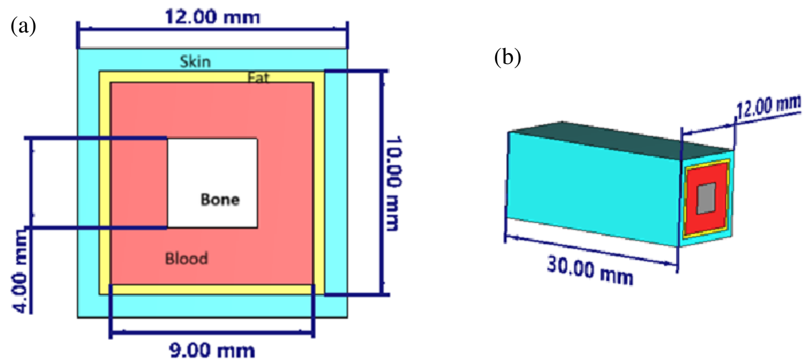

FIGURE 4. Finger phantom model: (a) top view and (b) isometric view.

TABLE 2. Dielectric constant values of blood for different glucose concentration.

Glucose (mg/dL)	ε_r of blood at 2.8 GHz
0	69
250	67
500	65

where ε^* represents the complex permittivity, ε_s the static permittivity, ε_∞ the permittivity at high frequency, τ the relaxation time(s), α the Cole-Cole dispersion parameter $(0 \leq \alpha \leq 1)$, σ the conductivity (S/m), and ω the angular frequency.

In the next step, it is important to determine the position of the finger over the antenna because it greatly affects the results. Fig. 5 illustrates the position of the FPM on the proposed antenna. It is placed at the edge of the square patch along the x-axis, where the E-field strength is at its maximum. Because the proposed antenna sensors are dual-polarized, the finger can be oriented along the y-axis as well. For simplicity, only one position was fixed, and all results were analyzed using one reference position, namely along the x-axis, to avoid measurement errors.

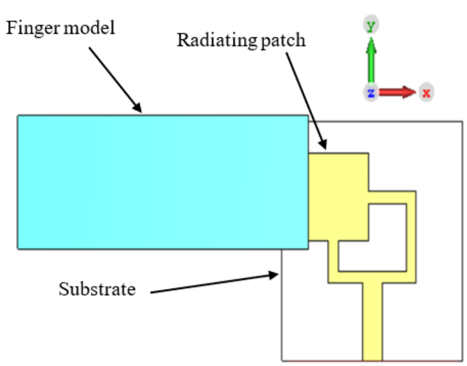

FIGURE 5. Positioning of the finger phantom on the proposed antenna.

The resonant characteristics of the antenna show shifts in their values with different blood glucose levels. The return loss vs. frequency plots for the two antenna sensors when the FPM is placed on it and the glucose is varied, as shown in Fig. 6, which demonstrates that the resonance frequency increases with increasing glucose concentration in the blood from 0 mg/dL to 500 mg/dL. This response shows how the presence of glucose alters the dielectric environment of the antenna, thereby affecting its resonant characteristics.

A plot of the jeans-based antenna (Fig. 6(a)) shows the distinct separation between glucose levels, indicating a high sensitivity and sharp resonance. The maximum frequency shift obtained was 78 MHz for glucose variations ranging from 0 to 500 mg/dL.

The second plot (Fig. 6(b)) confirms the moderate response of the FR4 based antenna. The maximum frequency shift achieved in this case was 57 MHz for glucose variations from 0 mg/dL to 500 mg/dL. In the case of a high dielectric material, such as FR4, the E-field is more confined to the substrate, lead-

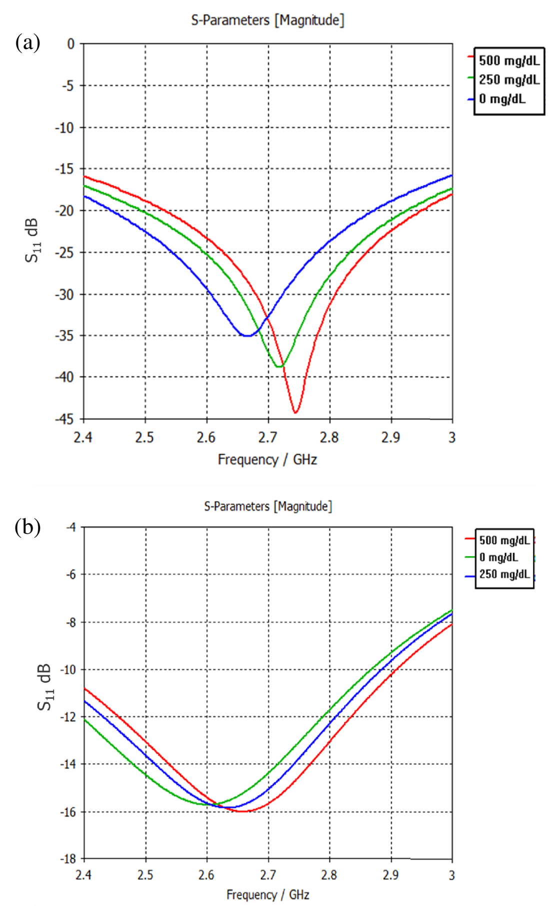

FIGURE 6. Resonance frequency shift with glucose concentration for (a) jeans-based and (b) FR4-based antenna.

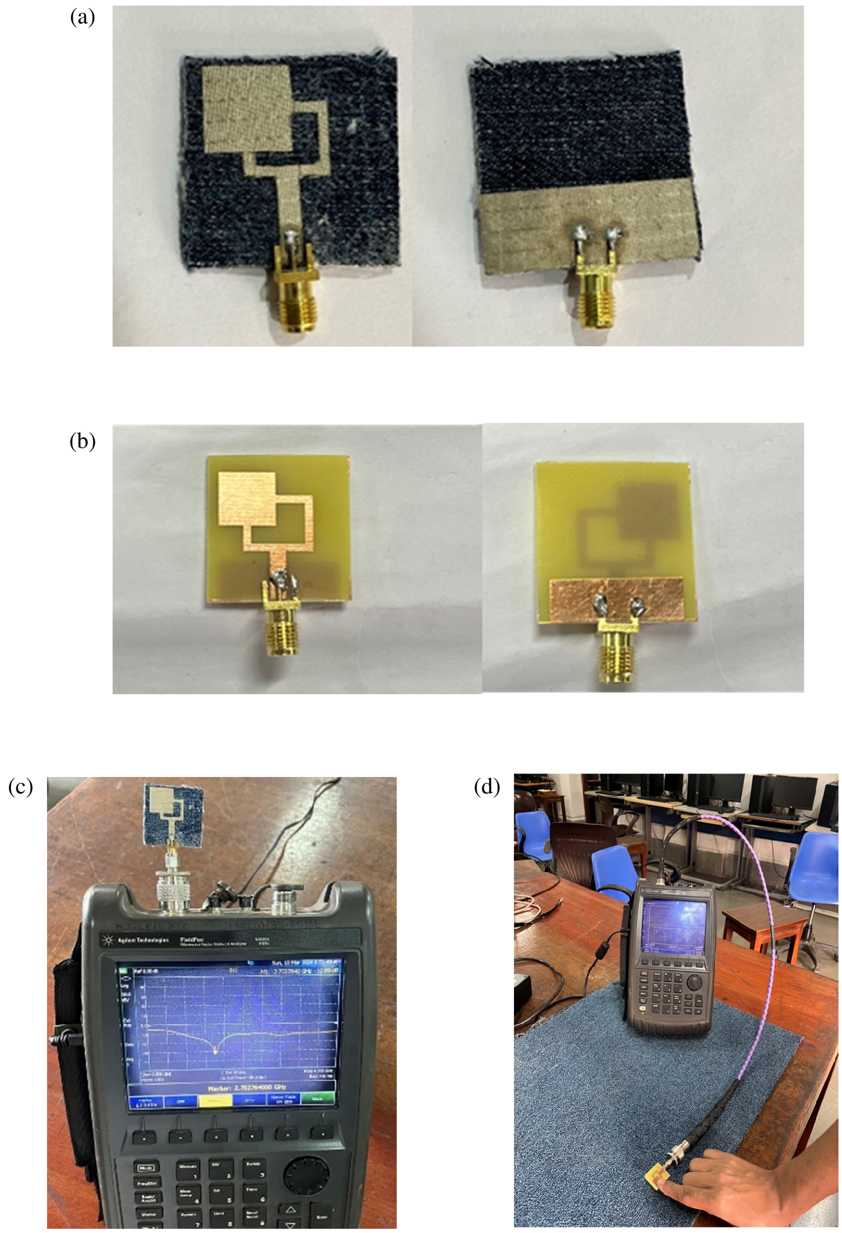

FIGURE 7. Photographs of the fabricated antennas: (a) jeans substrate, (b) FR4 substrate, and (c) antenna testing in free space, (d) antenna testing with finger loading.

ing to lesser tissue penetration and lower sensitivity. A low-dielectric material, such as jeans, enables deeper electromagnetic field penetration into the skin for better interaction with blood.

3.2. Prototype Fabrication and Free-Space Performance

The proposed jeans-based antenna was fabricated using a cutter and fabric fusing tape. A conductive fabric patch was pasted on the jeans using hot iron and tape. A 50 Ω SubMiniature version A (SMA) connector was attached carefully on low heat. The top and bottom views of the fabricated jeans-based antenna are shown in Fig. 7(a). The fabricated FR4-based antenna is shown in Fig. 7(b). The antenna resonant frequency was measured with the help of the Vector Network Analyzer (VNA), which was calibrated beforehand to avoid measurement errors. Fig. 7(c) shows the proposed jeans-based antenna tested on the VNA in free space, and Fig. 7(d) shows a practical experiment in which a finger is placed on the edge of the antenna, similar to the simulation environment conditions.

The measured resonant frequency in free space is 2.74 GHz as compared to simulated 2.8 GHz in the case of the jeans-based antenna (refer to Fig. 8). This shift is due to either fabrication errors or soldering losses, and can be expected from textile antennas fabricated manually, leading to human errors. The S_{11} value is -26.7 dB at 2.7 GHz showing good matching. In the case of the FR4-based antenna, the simulated and measured frequencies are well matched at 2.8 GHz, and the measured S_{11} is approximately -25 dB, as shown in Fig. 9.

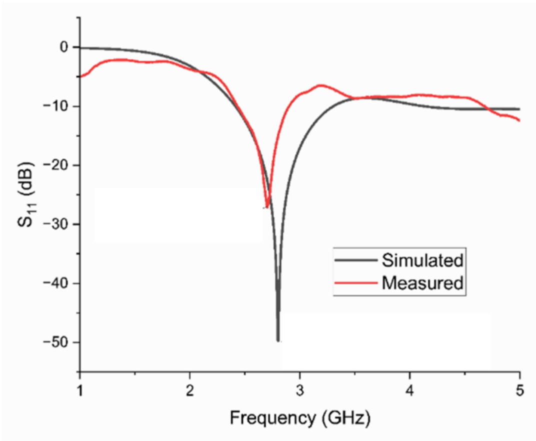

FIGURE 8. Simulated and measured return losses versus frequency for the jeans-based antenna in free space.

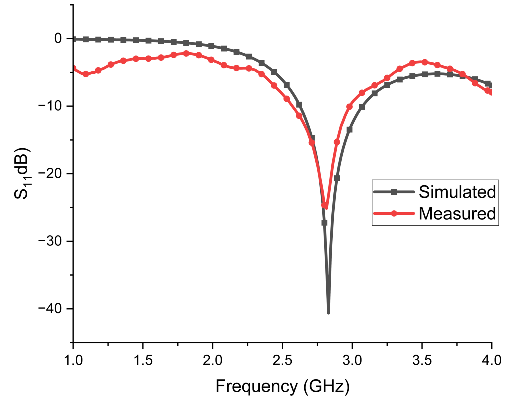

FIGURE 9. Simulated and measured return losses versus frequency for the FR4-based antenna in free space.

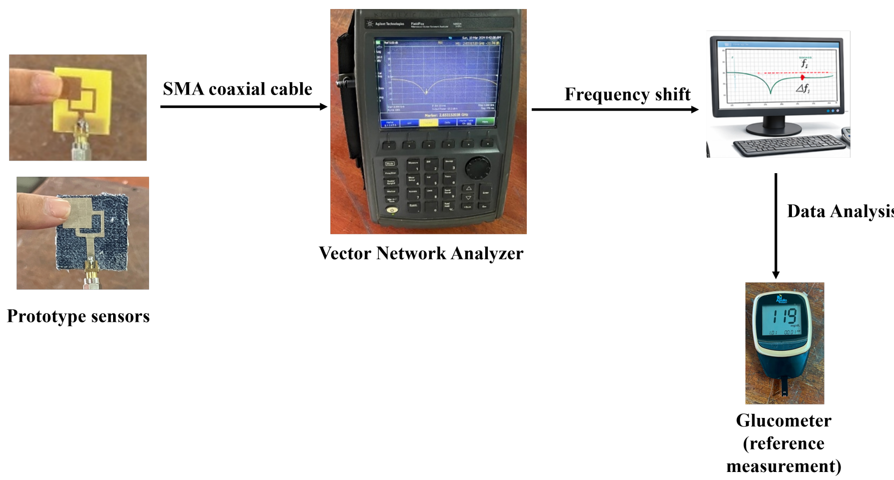

FIGURE 10. Experimental setup for glucose sensing.

3.3. Experimental Setup and Blood Glucose Monitoring Results

The experimental setup consisted of a calibrated vector network analyzer, fabricated antenna sensors, human volunteers, and a glucometer as depicted in Fig. 10. The volunteers were instructed to place their index finger on the edge of the patch antenna sensor along the x-axis to ensure measurement repeatability. Five samples were evaluated, and the glucose values of the volunteers were obtained using an invasive approach, that is, using a standard glucometer device. The finger was kept stationary, and the corresponding shift in the resonant frequency of the antenna was noted. The simulated and measured results obtained from these experiments were compared and discussed further.

The graph in Fig. 11 illustrates the variation in the resonant frequency with glucose concentration (mg/dL) for both the simulated and measured conditions for FR4-based antenna. The experiment shows that the resonant frequency shifts with glucose levels, indicating that the proposed antenna is sensitive to the variations in glucose levels of different volunteers, and the proposed design is suitable for noninvasive blood glucose sensing applications.

TABLE 3. Summarized results for FR4 based antenna.

Blood sugar (mg/dL)	Freq. (GHz) Sim.	S_{11} (dB) Sim.	Freq. (GHz) Measured	S_{11} (dB) Measured
119	2.6539	−16.13	2.6531	−31.74
125	2.6542	−16.01	2.6539	−21.46
144	2.6558	−15.47	2.6582	−25.35
147	2.6581	−16.54	2.6596	−22.00
152	2.6631	−16.25	2.6634	−28.05

From the results summarized in Table 3, the glucose values obtained from the volunteers ranged from 119 mg/dL to 152 mg/dL. The blood dielectric constant values corresponding to these measured glucose ranges were substituted into the phantom model design, and the antenna was re-simulated to determine the frequency.

The simulated frequency associated with the measured glucose values varies between 2.6539 GHz and 2.6631 GHz, with reflection coefficients in the range of −15.47 dB to −16.54 dB.

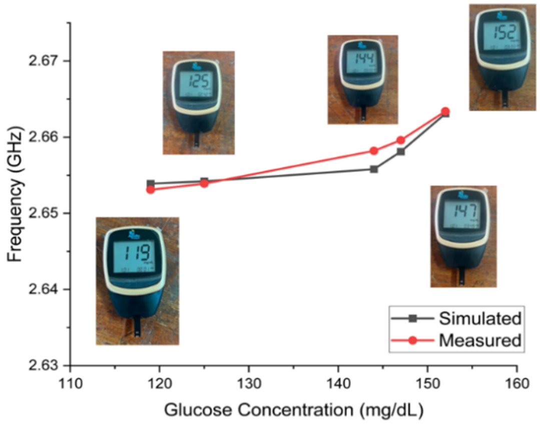

FIGURE 11. Simulated and measured resonance frequency shifts with glucose concentration for the FR4-based antenna sensor.

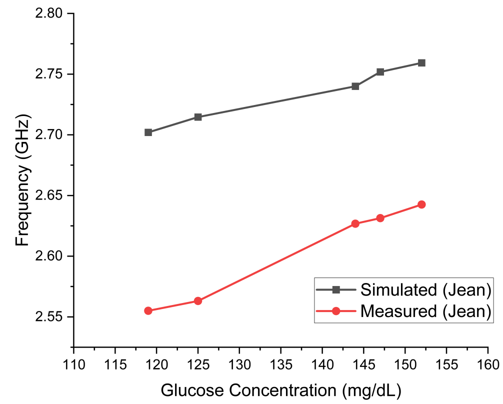

FIGURE 12. Simulated and measured resonance frequency shifts with glucose concentration for the jeans-based antenna sensor.

The measured data are in close agreement with the simulated ones, following the same trend, with resonance frequencies varying from 2.6531 GHz to 2.6634 GHz. Remarkably, the measured reflection coefficients lie between −21.46 dB and −31.74 dB, which is better than simulated values. When a human finger is placed on the patch antenna sensor, the patch experiences a change in the effective dielectric constant due to the high permittivity and lossy nature of biological tissues. In simulations, the simplified finger model with limited size does not fully represent the complex behavior of biological tissues. In real-time measurements, the effective tissue volume in close proximity to the antenna is larger, as the finger is part of the hand, which increases the electrical length of the antenna and enhances the dielectric loading effect, resulting in improved impedance matching as compared to simulations. The linear increase in resonance frequency with glucose concentration for both simulated and measured scenarios suggests that the proposed antenna sensor is reliable for glucose-sensing applications.

Similarly, measurements were performed for the jeans-based antenna. As the glucose concentration increases from 119 mg/dL to 152 mg/dL, both the simulated and measured resonant frequencies increased, as shown in Fig. 12, validating the sensitivity of the proposed antenna to glucose changes in blood.

Table 4 summarizes the simulated and measured results for the jeans-based antenna. The resonance frequency increases from 2.7020 GHz to 2.7593 GHz for glucose ranging from 119 mg/dL to 152 mg/dL. The simulated reflection coefficients exhibited good impedance matching and were below −31 dB across the glucose range. A similar frequency shift from 2.555 GHz at 119 mg/dL to 2.646 GHz at 152 mg/dL was observed in the measured results. The measured frequencies are lower than the simulated values and can be attributed to the fabrication tolerances. As mentioned earlier, the measured frequency for the jeans-based antenna was 2.74 GHz instead of 2.8 GHz, and hence, all other frequency shifts were eventually lowered. Although the measured frequencies showed minor discrepancies, the overall trend of the increase in frequency with glucose was clearly observed.

In summary, the jeans-based antenna sensor performed better than the FR4-based antenna in terms of impedance matching, bandwidth, gain, glucose sensitivity, and cost, as shown in Table 5. The gain of the antenna sensor impacts the electromagnetic field penetration and sensitivity. A high-gain antenna concentrates more electromagnetic energy toward the tissue, that is, in the intended direction, resulting in improved coupling efficiency and better frequency shift detection. On the other hand, a

TABLE 4. Summarized results for jeans-based antenna.

Blood Sugar (mg/dL)	Freq. (GHz) Sim.	S_{11} (dB) Sim.	Freq. (GHz) Measured	S_{11} (dB) Measured
119	2.7020	−32.51	2.555	−37.34
125	2.7146	−34.9	2.5631	−32.68
144	2.7402	−43.4	2.6268	−36.04
147	2.7518	−42.43	2.6314	−23.77
152	2.7593	−36.87	2.6426	−32.50

TABLE 5. Comparison between proposed jeans-based and FR4 antenna sensors.

Parameter	Jeans based antenna	FR4 based antenna
Resonant Frequency	2.8 GHz	2.8 GHz
Simulated Return Loss in free space	−50 dB	−42.38 dB
Bandwidth in free space	900 MHz (2.39–3.29 GHz)	460 MHz (2.59–3.05 GHz)
Gain	3.2 dBi	2.3 dBi
Glucose Sensitivity	High	Moderate
Fabrication Ease	High	Moderate
Cost	Very Low	Low

TABLE 6. Comparison of the proposed antenna with existing antennas for noninvasive glucose sensing.

Reference	Frequency (GHz)	Bandwidth (MHz)	Gain (dBi)	Maximum Frequency Shift (MHz)	Frequency Deviation (%)	Antenna size (mm^3)	Substrate
[8]	6.1	1900	3.3	800	13.11	$30 \times 30 \times 1.6$	FR4
[9]	5	35	NR	24	0.48	$20 \times 30 \times 1.6$	FR4
[10]	2.45	35	NR	5	0.20	$35 \times 13.5 \times 1.6$	FR4
[11]	5.5	5200	6.74	26	0.47	$30 \times 40 \times 1.6$	FR4
[12]	25.55	5.2	NR	100	0.39	$20 \times 15 \times 1.524$	FR4
Proposed Antenna 1	2.8	900	3.2	78	2.79	$35 \times 35 \times 1.25$	Jeans
Proposed Antenna 2	2.8	460	2.3	57	2.04	$27 \times 25 \times 1.6$	FR4

NR– data not reported

low-gain antenna radiates energy in multiple directions, thereby reducing field coupling with the tissue.

This study highlights the potential of low-dielectric substrates, such as jeans, as compared to high-dielectric substrates for noninvasive blood glucose sensing applications.

The antenna sensors in this study were compared with previous antennas in the literature for noninvasive glucose sensing applications and are summarized in Table 6. The frequency deviation is presented in terms of percentage using the following equation

$$\text{Frequency Deviation}\,(\%) = \frac{\Delta f}{f_0} \tag{4}$$

where Δf is the maximum frequency shift, and f_0 is the operating frequency of the antenna. The proposed antenna sensors exhibited a compact structure with good gain, sensitivity, and wide bandwidth. Compared to a few reported works [9–11], the proposed antennas showed better glucose sensitivity.

Overall, the jeans substrate shows a greater potential for sensing and user comfort, making it more suitable for continuous glucose monitoring applications.

For more accurate measurements, there is a need for a 3D enclosure that can securely hold the antenna and allow a finger to be placed precisely without applying excessive pressure on the antenna, because the textile material is flexible.

4. CONCLUSION

In this work, the design and analysis of a compact antenna sensor for noninvasive blood glucose sensing is presented for two different substrates. The Jeans substrate exhibited superior performance with a resonant frequency of 2.8 GHz, return loss of −50 dB, widest bandwidth of 900 MHz, and the highest gain of 3.2 dBi. It also offers high glucose sensitivity, is easy to fabricate, and has a very low cost, making it highly suitable for wearable, low-cost healthcare applications. The FR4 substrate showed moderate glucose sensitivity. Real-time testing using volunteers showed that jeans are highly sensitive substrates. The study shows that low-dielectric substrates improve field penetration into the skin, increasing the sensitivity to dielectric changes in blood. High-dielectric substrates tend to trap EM energy inside the substrate, reducing the sensitivity to glucose-induced dielectric shifts. The current study is based on a limited number of glucose samples, which restricts statistical reliability and can be considered as a proof-of-concept validation. In future work, the dataset will be expanded along with repeated measurements to improve the clinical reliability of the proposed antenna sensor.

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AUTHOR CONTRIBUTIONS

Monika Budania led the conceptualization, design, simulation, fabrication, and experimental validation of the proposed antenna and prepared the initial manuscript draft. Bharati Singh and Vandana Satam contributed substantially to the antenna design, technical validation, and the final review and approval of the manuscript.

DATA AVAILABILITY

The authors declare that all data supporting the findings of this work are available within the article.

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