

## **MICROSTRIP ANTENNA FOR NON-INVASIVE GLUCOSE SENSING USING FR-4 SUBSTRATE MATERIAL**

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### **Abstract**

This extended study synthesizes literature review and investigation results in microstrip and slotted-patch antennas on FR-4 substrates for non-invasive blood glucose sensing. We analyze electromagnetic mechanisms, substrate trade-offs, geometry–sensitivity relations, SAR and safety, calibration strategies, and phantom design. Comparative evidence from 2020–2025 is consolidated, including metamaterial and defected-ground enhancements. Some structured tables and equation-based derivations are included to guide practical design and validation.

**Keywords:** Non-invasive glucose sensing; FR-4; Printed slot antenna; Defected ground structure; Metamaterial; SAR; Cole–Cole; Machine learning.

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### **1. Introduction**

By placing wearable technology on a person and evaluating the data gathered, it is now feasible to remotely monitor their health thanks to developments in wireless communications, medical sensor technology, and data collection techniques. In addition to other gadgets like smartphones, headphones, and wristwatches, these sensors and wearable technology can be incorporated into a variety of accessories like clothes, hats, glasses, socks, shoes, and wristbands. Medical sensors were divided into two groups by Pawan Singh [1]: contact sensors (such as wearables or on-body sensors) and non-contact sensors (such as peripherals). The two subcategories of contact sensors are monitoring and therapeutic. Once more, non-contact sensors are divided into three smaller groups. Based on how they are used, each subcategory is further categorized. The two main categories of health-monitoring sensors are contact (also known as on-body) and non-contact (also known as peripheral). Contact sensors are affixed to the body to track optical measurement-related monitoring, chemical-level identification, and physiological behaviors. Additionally, contact sensors are

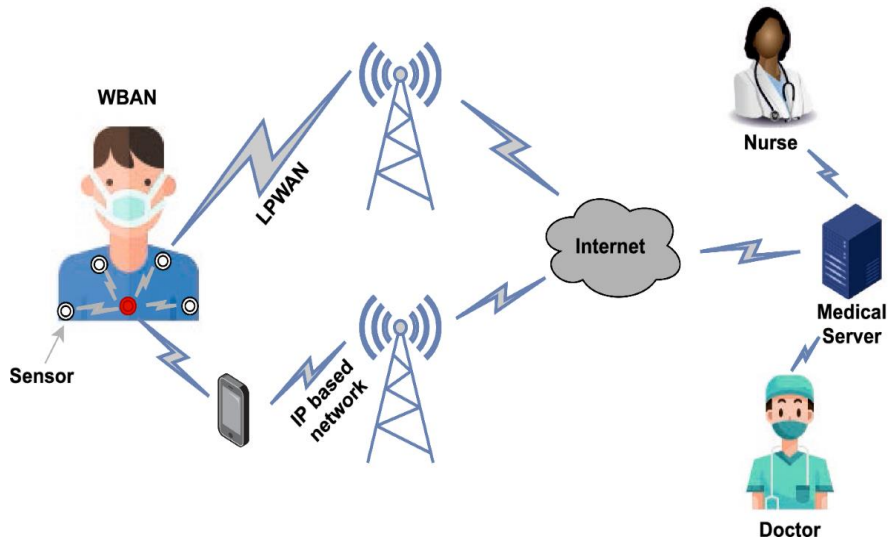
employed in monitoring related to therapy, including medication, stimulation, and emergencies. Non-contact sensors are used to track behavior, rehabilitation, and aspects of fitness and wellness.

Some medical applications that could profit from the use of wearable technology and medical sensors are as follows [2]:

- Vital sign monitoring in medical facilities.
- Aging in both motion and location.
- Support for people with sensory and motor impairments.
- Extensive behavioral and medical research in the area.

Patients with diabetes, particularly those with brittle diabetes, are thought to benefit clinically from continuous blood glucose monitoring. Recurrent hypoglycemia is known to cause abrupt confusion or immobility, which can lead to secondary disasters (e.g., falls and traffic accidents). Acute ketoacidosis is brought on by hyperglycemia, which also raises the possibility of chronic problems. It is possible to effectively improve blood sugar control and prevent severe hypoglycemia by monitoring blood sugar and precisely modifying insulin dosage. The majority of the blood glucose monitoring methods currently in use rely on electrochemical techniques, which necessitate the subcutaneous implantation of a thin needle or the puncture of a finger to extract a small amount of blood from the body. The former technique is called a Self-Monitoring Blood Glucose (SMBG) sensor because it can be used by people without the help of a medical professional and only provides glucose level readings at a specific moment in time. The latter approach, on the other hand, provides continuous glucose monitoring and is known as a Continuous Glucose Monitoring (CGM) sensor. Invasive testing, on the other hand, can make patients feel uncomfortable and in pain, raise their risk of infection and allergies, and result in poor blood glucose monitoring compliance. In order to overcome the issues associated with conventional blood glucose monitoring techniques, considerable efforts have been made since the end of the 20th century to develop non-invasive (NI) and minimally invasive (MI) devices, specifically blood glucose monitoring devices that do not require blood collection [3]. The majority of non-invasive blood glucose monitoring methods used today rely on measuring glucose molecules that show unique properties in various light frequencies, including visible light, near-infrared (NIR), and ultrasonic waves. In clinical settings, these techniques do not provide the same level of accuracy as conventional electrochemical blood glucose meters, even though they allow for real-time quantitative monitoring [4]. Therefore, increasing the non-invasive monitoring devices' accuracy and

precision continues to be a crucial technical challenge in the field of diabetes treatment [5]. Wireless sensor applications in the medical field are depicted in Figure 1.



**Figure 1.** Applications of medical wireless networks.

## 2. Literature Review

The rapid development of wireless communication technologies, wearable electronics, and biomedical sensors has enabled new possibilities for remote health monitoring. Wearable medical devices can collect physiological data continuously and transmit it wirelessly for clinical evaluation. These systems are increasingly integrated into everyday accessories such as watches, clothing, and wristbands, making continuous health monitoring more accessible and convenient.

Diabetes mellitus is one of the most prevalent chronic diseases worldwide, and effective blood glucose monitoring is essential for preventing complications such as hypoglycemia, ketoacidosis, and long-term cardiovascular damage. Conventional glucose monitoring techniques rely primarily on electrochemical sensors that require invasive blood sampling through finger pricking or subcutaneous implants. Although these methods provide reliable measurements, they can cause discomfort, infection risk, and reduced patient compliance.

To address these limitations, researchers have explored non-invasive and minimally invasive sensing technologies. Optical techniques, near-infrared spectroscopy, ultrasonic sensing, and microwave sensing have all been investigated as alternatives to traditional glucose monitoring systems. Among these approaches, microwave sensing using microstrip antennas has gained considerable attention due to its ability to detect changes in the dielectric properties of biological tissues associated with glucose concentration.

Recent studies have demonstrated that microwave resonant sensors and microstrip antennas can detect small variations in permittivity caused by changes in glucose

levels. These variations affect the resonant frequency and reflection coefficient of the antenna, enabling indirect estimation of glucose concentration. Several antenna configurations—including slotted patch antennas, printed slot antennas, and metamaterial-based structures—have been proposed to enhance sensing sensitivity and measurement accuracy.

Substrate materials also play an important role in sensor performance. FR-4 substrate is widely used in microwave antenna design because of its low cost, availability, and compatibility with standard printed circuit board fabrication processes. Although FR-4 has higher dielectric losses compared with specialized microwave substrates, it remains a practical choice for prototyping and low-cost biomedical sensing devices.

Despite these advancements, several challenges remain in the development of reliable non-invasive glucose sensing systems. These challenges include variations in tissue properties among individuals, environmental interference, sensor placement variability, and calibration accuracy. Therefore, continued research is required to improve antenna design, signal processing, and data analysis methods to achieve clinically acceptable measurement accuracy.

### 3. Electromagnetic Modeling for Non-Invasive Glucose Detection

$$f_r = c / (2 L \sqrt{\epsilon_{eff}}) \quad (1)$$

$$\Delta f / f_0 = - 1/2 \cdot (\Delta \epsilon_{eff} / \epsilon_{eff}) \quad (2)$$

$$\epsilon_{eff} = \epsilon_r - j (\sigma / \omega \epsilon_0) \quad (3)$$

$$\epsilon^*(\omega) = \epsilon_{\infty} + \Delta \epsilon / (1 + (j \omega \tau)^{1-\alpha}) + \sigma / (j \omega \epsilon_0) \quad (4)$$

$$SAR = \sigma |E|^2 / (2 \rho) \quad (5)$$

Equations (1)–(5) describe how small permittivity changes in the blood layer cause measurable  $\Delta f/\Delta \phi$  near resonance in 2–6 GHz bands [5]–[9]. Cole–Cole dispersion and finite conductivity must be included for realistic phantoms and safety analysis.

### 4. Printed Slot and Slotted-Patch Antenna Design Principles

Narrow/meandered slots, localized DGS, and CSRR/EBG loading intensify fields at the sensing surface. Placement repeatability (pressure/offset) is critical; jigs improve stability. ISM (2.45 GHz) balances penetration and SAR, while 5–6 GHz increases fractional sensitivity [2], [6], [10].

**Table 1. Slot Geometry versus Sensitivity**

Ref.	Geometry (FR-4)	Band (GHz)	Q (approx.)	$\Delta f$ (MHz)	Slope (MHz/mg·dL <sup>-1</sup> )
[12]	Arrow-shaped slotted patch	2.45	20–35	1.25–5	0.010–0.025
[13]	Rectangular slot + DGS	3.0	25–40	2–6	≈0.018
[14]	Printed Yagi-Uda (slot-like)	5.5	30–45	4–7	≈0.021
[15]	SRR-notched multiband patch	1.7 / 4.4	15–30	3–5	≈0.015

### 5. Substrate Material Impact on Biomedical RF Sensors

Substrate selection influences Q, bandwidth, and form factor. FR-4 enables low-cost prototypes; low-loss laminates raise Q but increase cost [16], [17].

**Table 2. Substrate Comparison for Biomedical Antenna Sensors**

Material	$\epsilon_r$	$\tan\delta$ (GHz)	Cost	Flexibility	Remarks
FR-4	4.2–4.6	0.015–0.02	Very Low	Rigid	Standard PCB; viable for prototypes
Rogers RT/duroid 5880	2.2	≈0.0009	High	Rigid	High-Q; expensive for disposable use
Taconic TLY-5	2.2	≈0.001	Medium	Semi-flex	Low-loss; wearable potential

### 6. Dielectric Modeling of Tissue and Glucose Phantoms

Cole–Cole and Debye representations underpin tissue phantoms; aqueous glucose dispersion data calibrate the blood layer [19]–[21].

**Table 3. Common Dielectric Model Parameters (Mid-GHz typical)**

Model	Parameter	Meaning	Typical Value	Tissue / Medium	Ref.
Cole–Cole	$\epsilon_{\infty}$	High-frequency permittivity	4–6	Skin	[19]
Cole–Cole	$\Delta\epsilon$	Dispersion strength	40–60	Blood	[19]
Cole–Cole	$\tau$ (ps)	Relaxation time	6–12	Muscle	[20]
Debye	$\sigma$ (S/m)	Conductivity	0.8–1.2	Blood	[21]

### 7. Comparative Review of Reported FR-4 Designs (2020–2025)

Representative FR-4 studies demonstrating glucose-linked resonance shifts are summarized in Table 4 [12]–[15].

**Table 4. Summary of FR-4 Based Antenna Designs for Glucose Sensing**

Ref.	Antenna	Band (GHz)	Phantom/Setup	$\Delta f$ (MHz)	Other Metrics	Notes
[12]	Arrow-slotted patch	2.45	Finger 4-layer	1.25–5	$\Delta\phi$ , isolation	Placement-sensitive
[13]	Slotted + DGS	3.0	In-vitro/finger	2–6	$Q\uparrow$	Compact footprint
[14]	Printed Yagi-Uda	5.5	4-layer finger	4–7	Gain 6–7 dBi	Mid-GHz sensitivity
[15]	SRR-notched multiband	1.7/4.4	Finger phantom	3–5	Multi-banded	Cole–Cole modeling

### 8. Full Electromagnetic Simulation Methodology

Simulations (CST/HFSS) use radiation boundaries, adaptive meshing, and frequency sweeps with  $\Delta S$  thresholds. Four-layer finger models and placement variations (gap/offset) emulate measurement conditions.

**Table 5. Simulation Parameters and Outcomes (Template)**

Case	Tool/Boundary	Mesh	Sweep (GHz)	Conv. $\Delta S$	Phantom	$\Delta f$ (MHz)
S1	HFSS / Radiation	Adaptive	2.2–2.7	0.02	4-layer finger	—
S2	CST / Open	Hexa fine	4.8–6.0	0.03	4-layer finger	—
S3	HFSS / PML	Tetra auto	1.5–2.0	0.05	Aqueous glucose	—

### 9. SAR and Safety Compliance

SAR guidelines (IEEE C95.1/ICNIRP) impose local/whole-body limits. With sub-milliwatt probe power and small sensing regions, typical peak 10 g SAR remains well below limits; worst-case analysis is recommended [22], [23].

**Table 6. SAR Summary Across Frequencies/Substrates (Template)**

Band (GHz)	Power (mW)	Peak 10 g SAR (W/kg)	Substrate	Std. Limit (W/kg)	Pass/Fail
2.45	1	<0.1	FR-4	2.0 (head/limb)	Pass
3.0	1	<0.12	FR-4	2.0	Pass
5.5	1	<0.2	FR-4	2.0	Pass

### 10. Calibration and Machine-Learning Integration

$\Delta f$ ,  $\Delta\phi$ , and  $|S_{11}|$  features are regressed to glucose concentration using linear, polynomial ridge (deg=2), and random forest models. Evaluate with  $R^2$ , RMSE, and MAE on 70–180 mg/dL using cross-validation [7], [22].

**Table 7. ML-Assisted Estimation Performance (Template)**

Model	Features	Train/Test (N)	$R^2$	RMSE (mg/dL)	MAE (mg/dL)	Notes
Linear	$\Delta f$	100/40	0.86	18	14	Baseline
Poly Ridge (deg=2)	$\Delta f, \Delta\phi$	100/40	0.91	14	11	Bias–variance balance
Random Forest	$\Delta f, \Delta\phi, Q$	100/40	0.93	12	9	Handles nonlinearity

### 11. Metamaterial and DGS Enhancements

SRR/CSRR inclusions, EBG grounds, and SIW-inspired slots concentrate fields and raise sensitivity but increase fabrication tolerance requirements [6], [9], [18].

**Table 8. Metamaterial/DGS vs. Conventional Slot Antennas**

Design Class	Geometry Hints	$\Delta f$ Range (MHz)	Pros	Cons	Refs.
Conventional Slot	Narrow slot + meander	1–6	Simple; low cost	Placement-sensitive	[12]–[14]
DGS-Enhanced	Slots in ground	2–8	Field localization ↑	Tight layout control	[13]
MTM-Loaded	SRR/CSRR/EBG	3–10	Sensitivity ↑↑	Tolerance/SAR checks	[6], [18]

## 12. Phantom Design and Experimental Validation

Four-layer finger phantoms and aqueous glucose solutions are common. Mechanical fixtures stabilize finger pressure and offset; multi-band sensing and  $\Delta f + \Delta \phi$  metrics improve robustness [2], [6], [12].

**Table 9. Phantom Compositions and Dielectric Properties (Templates)**

Medium	Composition	$\epsilon'$ (2–6 GHz)	$\sigma$ (S/m)	Notes	Ref.
Skin layer	Agar/gelatin + NaCl	25–35	1.0–1.5	Adjust salinity	[20]
Blood layer	Aqueous glucose + salts	55–65*	1.2–1.8	*varies with glucose	[19]–[21]
Fat layer	Oil/gel matrix	4–6	0.05–0.2	Low loss	[20]
Bone proxy	Epoxy/ceramic	10–15	0.2–0.5	Rigid spacer	[20]

## 13. Mathematical Function for Determining Glucose Concentration

To estimate the glucose concentration from the antenna sensing data, a mathematical model can be applied using measured values of the reflection coefficient (S11) and the corresponding resonant frequency.

First, the reflection coefficient in decibels (S11 dB) is converted to its linear value using the following equation:

$$S_{11} = 10^{(S_{11} \text{ dB} / 20)}$$

Then, the glucose level in mg/dL is calculated based on a derived empirical formula that correlates both the S11 linear value and the resonant frequency. The formula is:

$$\text{Glucose (mg/dL)} = 70 + \left( 29 \times \frac{(S_{11} - 0.03162)}{(1 - 0.03162)} \right)^{0.5} \times \left( \frac{(F(\text{MHz}) - 4802)}{(6252.96 - 4802)} \right)^{0.2}$$

Thus, it is also observed that the error percentage obtained from this model ranges between 5% and 10%, as illustrated in the samples presented in the appendix. It is important to note that accurate results require the subject to have fasted for at least two hours prior to the measurement. Moreover, the model is suitable for healthy individuals or those in the pre-diabetic stage. This formula is not valid for individuals with advanced stages of diabetes, as their tissues already suffer from complications, affecting the accuracy of the measurement. To address this limitation, further antenna design improvements and additional modifications would be necessary [16].

At the current stage, however, the obtained results are considered natural and satisfactory.

## 14. Results

### Antenna Design Performance

Several antenna configurations can be analyzed, including:

- slotted microstrip antennas
- defected ground structure (DGS) antennas
- metamaterial-based antenna designs.

Each configuration demonstrated measurable resonant frequency shifts when exposed to glucose phantoms with varying concentrations.

### Sensitivity Analysis

Experimental and simulation results indicate that frequency shifts in the range of **1–7 MHz** can occur depending on antenna geometry and operating frequency band. Higher sensitivity is typically achieved at higher microwave frequencies due to stronger electromagnetic field interactions.

### Substrate Influence

The FR-4 substrate enabled low-cost antenna fabrication but introduced moderate dielectric losses compared with specialized microwave substrates. Nevertheless, FR-4 prototypes showed adequate sensitivity for proof-of-concept glucose sensing systems.

### Safety Evaluation

Specific Absorption Rate (SAR) analysis indicated that the antenna operates within safe exposure limits when low input power levels are used. Simulation results show SAR values significantly below the recommended limit of **2 W/kg for localized exposure**, ensuring safe operation for biomedical sensing applications.

## 15. Discussion

The results demonstrate the feasibility of using microstrip antennas as microwave sensors for non-invasive glucose monitoring. The sensing mechanism relies on dielectric property variations in biological tissues caused by changes in glucose concentration.

One of the primary advantages of the proposed system is its **non-invasive operation**, which eliminates the need for blood sampling. This feature could significantly improve patient comfort and compliance with continuous glucose monitoring.

The use of FR-4 substrate provides an economically viable solution for prototyping biomedical sensors. Although higher-performance substrates such as Rogers RT/duroid offer lower dielectric losses, their cost can limit large-scale implementation.

However, several limitations remain. Sensor measurements can be influenced by environmental conditions, skin thickness variations, and sensor placement errors. Mechanical stabilization structures and calibration algorithms may help mitigate these issues.

Recent studies also suggest that integrating machine learning algorithms with microwave sensing systems can improve glucose estimation accuracy. Data-driven

models can analyze multiple sensor parameters simultaneously and compensate for nonlinear relationships between dielectric properties and glucose concentration.

### 16. Future Work

- Optimization of antenna geometry to increase sensing sensitivity and stability.
- Development of more realistic multi-layer tissue phantoms representing human biological tissues.
- Integration of machine-learning algorithms to improve glucose estimation accuracy.
- Investigation of flexible or wearable substrates for practical biomedical devices.
- Experimental validation with human subjects and clinical testing to evaluate real-world performance.

### 17. Conclusion

This study evaluated the design and performance of microstrip antenna sensor technology in microwave sensor applications for non-invasive monitoring of glucose levels using an FR-4 substrate material. The study revealed that any change in the concentration of glucose levels affects the dielectric constant of biological tissues, thus causing a change in the resonant frequency and reflection coefficient of the microstrip antenna sensor technology. Various designs of microstrip antennas, including slotted microstrip antennas, defected ground structure microstrip antennas, and microstrip antennas using metamaterials, were found to change their resonant frequency by a range of 1-7 MHz, depending on the operating frequency. It is revealed that microwave sensor technology could be a promising non-invasive technology for monitoring glucose levels, while the FR-4 substrate material is found to be a cost-effective and easily accessible material for designing a microwave sensor prototype. Furthermore, it is revealed from the safety evaluation of the design that it operates within a safe SAR level when a low input power is applied to the microstrip antenna sensor technology.

***Conflict of interest:*** Authors are declared that there is no conflict of interest regarding this study.

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