A Wireless and Battery-free Biosensor Based on Parallel Resonators for Monitoring a Wide Range of Biosignals

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Abstract—This paper proposes a novel wireless, battery-free, and label-free biosensor for minimally invasive and non-invasive permittivity sensing for applications such as detecting glucose levels in the interstitial dermal fluid. The miniaturized, fully passive sensor is based on two symmetric parallel 0.8 mm³ LC (inductor-capacitor) resonators. Each inductor is integrated with a conductive plate at one of its terminals. The passive sensor’s main field is dominant outside the resonators, creating an effective capacitance outside the sensor, unlike reported in previous studies where the effective capacitance was limited to between the resonator lines. The proposed sensor was further analyzed under two different scenarios to change the region of effective capacitance and test its suitability for pressure monitoring, such as wound monitoring; first, by repositioning one of the two resonators, and second, by replacing one of the resonators with a conductive plate. The experimental results confirmed the proposed passive sensor’s performance in detecting the glucose concentration in an aqueous solution with a sensitivity of 500 and 46 kHz/(mg/dL) for minimal invasive and non-invasive monitoring, respectively, within the glucose range of 0-500 mg/dL with volume sample requirements as low as 60 μL. Further, by repositioning one of the resonators, the effective capacitance lies inside the passive sensor, making it suitable for wound monitoring. The results indicated that the resonance shift can be made fairly linear with respect to the variation in the separation between the resonators with a sensitivity of 2.5×10³ MHz/mm within the separation range of 0.2-0.5 mm, considering separation as the main factor to sense the pressure or healing of the wound.

Index Terms—Sensor telemetry, circuit analysis, minimally invasive, non-invasive, passive LC resonators, pressure monitoring, permittivity, blood glucose monitoring.

I. INTRODUCTION

MICROWAVE spectroscopy is a powerful tool for monitoring anomalies in biological fluids that can detect a type of cancer, burn depth, and blood sugar levels [1]–[8]. Microwave spectroscopy is label-free, non-intrusive, antibody-free, and uses non ionizing radiation, which makes it highly appropriate for biomedical applications. Biological tissues, like any other material, exhibit unique dielectric properties and electric polarization (P), described as permittivity, which determines the degree of interaction of electrical polarization of a material with the applied external electric field [9]. Any variation in the permittivity of biological tissues mainly occurs due to a change in its electric polarization, therefore indicating an anomaly. Thus, microwave spectroscopy is one of the promising diagnosis techniques.

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In general, microwave spectroscopy techniques can be classified into resonant and non-resonant techniques for the characterization of biological tissues. The resonator technique is most commonly used in biomedical applications because it accurately indicates changes observed in biological tissues. The resonator-based biosensor is based on the interaction of the biosignals with either the main or fringe field. Although the fringe field can be used to monitor changes in the biosignals [10]–[12], it has a low sensitivity in comparison with that of the main field; and hence, the main field is preferred for monitoring of biosignals [13]. Recently, many wired active resonator-based biosensors have been designed to confine the electric fields between resonator lines in a specific region with high potential difference, by using devices, such as an interdigital capacitor [14]–[16] or a split ring resonator [17]–[20], provided the biological tissues are dropped in that specific region. Thus, to facilitate the use of these biosensors in wearable applications, the biosignals interact with the fringe fields, while the main field is located between the resonator lines. As a result, the sensitivity of the biosensors reduces because the biosignals are not located between the resonator lines. Thus, a paradigm shift is required to acquire a strong electric field outside the resonator without locating biosignals between the resonator lines. The above issue is addressed in this study by designing a novel biosensor based on two symmetric parallel LC resonators, where the strength of the

Fig. 1. Schematic of the fully passive resonator-based biosensor for wearable applications (i.e., position A) or minimally invasive (i.e., position B) (a) Glucose monitoring (b) Wound monitoring
electric field is dominant outside the biosensor, unlike in the previous studies in which the main field was confined between the resonator lines [10]–[22]. Additionally, we used a wireless, battery-free resonator-based biosensor to monitor the biological tissues owing to its compact size, light weight, cost efficient fabrication, and convenient use in wearable and implantable applications [23].

Over the last decade, wireless, battery-free sensing systems have gained significant attention in the biomedical field. The system comprises a passive resonator that interacts with biological tissues, and an external reader that tracks the resonance frequency of the passive resonator outside the tissues. Thus, many studies have focused on improving either the passive resonator or the external reader. The application of microelectromechanical system (MEMS) technologies has been widely introduced for improving the structure and performance of passive resonators. This system monitors the continuous wireless physiological parameters for various types of pressure by using the passive resonator [25], and focuses on using passive resonator comprising of advanced materials such as air cavities [29]–[31], flexible pyramidal dielectric materials [29]–[31], or flexible sponge materials [32] to improve the performance of the resonator. The passive resonator-based pressure monitoring used in these studies are based on the parallel-plate technique, with either an inductor on the top of the plate, [26]–[30], or using two parallel inductors [31], [32]. Furthermore, based on wireless power transfer technologies, improving the wireless power transfer to the passive resonator improves the ability of the external reader to track the resonant frequency of the passive resonator. For example, replacing a vector network analyzer with a portable reader has been introduced [33]. Moreover, systems, such as the parity-time (PT) symmetry and exceptional points (EPs), have also been studied for improving wireless power transfer to the passive resonator [25], and focuses on using passive resonator based biosensors to enhance the sensing capability of the biological tissues [34], [35]. However, none of these studies have explained the reason and the advantages behind using the specified resonator structures owing to shortcomings in the analysis and equivalent circuit.

This paper raises a key question: Does any passive sensor based on parallel plates, where the plates are separated by a distance s, induce an effective capacitance inside the sensor? Considering this, the proposed passive sensor was further analyzed by changing one of the resonators, either by repositioning it or replacing it with a conductive plate to serve as three different passive sensors. Three passive sensors based on the parallel plate technique were studied. The direction of current through the inductor and the charge distribution were taken into consideration to investigate the direction of uniform electric field between the plates, and effective capacitance. Consequently, the sensing performance of the sensor was improved for both pressure sensing (i.e., wound monitoring) and permittivity sensing (i.e., glucose monitoring). To the best of our knowledge, this paper is the first to analyze the region of dominance of the electric field in various passive sensors. Therefore, the key contributions of this study include:

- A conceptual understanding of the region of dominance of the electric field in different passive sensors to monitor wide-range biosignals is presented. As this topic is largely ignored, we believe that it could be beneficial in designing passive resonator-based biosensors.
- A novel approach to analyze the glucose level in the fluid is developed. It can operate underneath the skin for a minimal invasive or non-invasive approach, as shown in Fig. 1(a).

II. SYSTEM ANALYSIS AND PRINCIPLE OF OPERATION

The proposed passive sensor is based on two symmetric parallel inductors, as shown in Fig. 2(a). Further, a change in any one of the inductors, such as repositioning or being replaced with a conductive plate, is studied, as shown in Fig. 2(b) and (c). Three miniaturized passive sensors were designed, simulated, and fabricated on a $2 \times 2 \times 0.2$ mm$^3$ plane flame retardant 4 (FR4) substrate. An FR4 has a relative permittivity of 4.4 and a loss tangent of 0.02. The passive sensors were designed using a 3D full-wave finite element simulator (ANSYS HFSS) and fabricated using standard printed circuit board technology. A single turn loop coil was used as external reader, which was placed outside the biological tissues, as shown in Fig. 3(a) and (b). Based on the near field (non-radiative technique) the passive sensors were inductively powered using the external reader.

Table I lists the parameters of the sensor’s inductor. The top and bottom inductor are symmetric; and hence, $L_1 = L_2 = L_s$, where $L_s$ represents the inductance of the passive sensor, and can be defined by:

$$L_s = \frac{2\mu_0}{\pi} n^2 d_{avg} \left[ \ln \left( \frac{2.07}{\phi} \right) + 0.18\phi + 0.13\phi^2 \right]$$

(1)

where $d_{avg} = \frac{d + d_{in}}{2}$ is the average diameter of the square, and $\phi = \frac{d_{in}}{d}$. $\alpha$ is the ratio of the outer diameter to the inner diameter [36], [37]. $\mu_0$ is the permeability of the free space.

Fig. 2. Top view of the passive sensor (a) proposed sensor: two parallel inductors turn clockwise with the same start point and current direction flow through both inductors (sensor 1). (b) Repositioning one of the inductors of proposed sensor: top inductor turns counter clockwise while the bottom inductor turns clockwise with opposite start point and, current direction flow through the inductors is opposite to each other (sensor 2). (c) Replacing one of the inductors of the proposed sensor with a conductive plate (sensor 3).
TABLE I. Geometrical parameters of the inductor

| Parameter | Description | Values (unit) |
|-----------|-------------|---------------|
| n         | Number of turns | 3             |
| e         | Line spacing  | 0.1 mm        |
| w         | Line width    | 0.1 mm        |
| t         | Line thickness| 0.017 mm      |
| p         | Inner plate   | 0.8 mm        |
| d         | Outer diameter| 2 mm          |

Table II. Characteristic frequencies of a planar spiral inductor

| Frequency (GHz) | Calculated in air | Simulated in air | Simulated on FR4 |
|-----------------|-------------------|------------------|------------------|
| f₀ = 5.1        | f₀cal = 5.7       | f₁ = 4.2         |

$\mu_0 = 4\pi \times 10^{-7}$ N.A$^{-2}$. The inductance of the passive sensor $L_w$ was approximately calculated as 20 nH. Furthermore, the distributed capacitance of a spiral inductor was studied in [38], [39]. Generally, for any pattern, the distributed capacitance is given as:

$$\frac{1}{C_0} = \frac{1}{4\pi \epsilon_0 Q^2} \int \int \left[ \frac{\rho(r_i)\rho(r_j)K_{ij}}{|r_i - r_j|} \right] d^3r_i d^3r_j (2)$$

The integration is applied over the entire area of the pattern. $Q$ is a total charge, and $\rho(r_i)$ is the spatial charge density as a function of $r_i$, which is the length of the inductor trace. $\epsilon_0$ is the permittivity of the free space ($\epsilon_0 = 8.85 \times 10^{-12}$ F.m$^{-1}$).

The distributed capacitance $C_0$ of the inductor of the passive sensor was approximately calculated as 0.05 pF [40]–[42]. Thus, from (1) and (2), the self-resonance frequency of the spiral inductor of the proposed sensor was calculated, where the resonance for the inductor in the free space was given as $f_0 = 1/2\pi\sqrt{C_0 L_w}$. Table II shows a comparison of the calculated resonant frequency and the simulation results. Fig. 3 shows the resonant frequency of the spiral inductor of the passive sensor through the external reader. Further, the frequency decreases to 4.2 GHz when the inductor is positioned at the top of FR4, considering a part of the fringe capacitance of the inductor lines interacts with FR4 [38]–[43].

Furthermore, the capacitive coupling and resonance of the passive resonators are studied based on the parallel plate technique. Additionally, the direction of current through the inductor depending on the external reader coil is analyzed to understand the charge distribution on the inductor, considering it could improve the sensitivity of the sensor towards the pressure and permittivity of applications.

A. Two-symmetric inductor with same current direction

Based on the parallel plate technique, the two symmetric spiral inductors of the proposed passive sensor have the same turn direction (clockwise), as shown in Fig. 2(a). Additionally, the current in both inductors also move in the same direction, therefore creating identically charged inductors. Thus, according to Gauss’s law, there is not enough potential difference between the two inductors, and the direction of the electric field is reversed to the outside the sensor. As a result, there is no effective capacitance between the two-symmetric inductors, which means no energy is stored inside the sensor. Fig. 2(a) illustrates the circuit analysis by applying the dot convention, which represents the turn direction of the inductors, which are adjacent to each other. The current leaving the dotted terminal of the top inductor is in-phase with the current leaving the dotted terminal of the bottom inductor, and hence the polarities of the voltages at both dotted terminals are also in-phase. Therefore, when the voltage is positive at the dotted terminal of the top inductor, the voltage at the dotted terminal of the bottom inductor is also positive. Fig. 2(b) demonstrates the uniform electric field between the inner surface of the inductors plates at region II (RII), which according to the principle of superposition given as:

$$\int \vec{E}_{in} \cdot d\vec{a} = E_{in}^{net} a$$

$$E_{in}^{net} = E_{in}^1 + E_{in}^2 = -\frac{Q_1}{\epsilon a} + \frac{Q_2}{\epsilon a} = 0$$

where a and $\epsilon$ represent the surface area of the plate and the permittivity at RII, respectively, and the net electric field between the induction lines of both inductors is found to be zero. However, the net uniform field outside the passive sensor (at the top and bottom sides) is dominant. The electric field of the inner surface of the inductors plates at the top, at region I (RI) can be represented as:

$$E_{out}^{net} = E_{out}^1 + E_{out}^2 = \frac{Q_1}{\epsilon a} + \frac{Q_2}{\epsilon a} = \frac{Q}{\epsilon a}$$

where the permittivity of free space $\epsilon_0$ is outside the passive sensor. Similarly, the electric field at the bottom side, at region III (RIII) is given as:

$$E_{out}^{net} = -\frac{Q}{\epsilon_0 a}$$

From the above analysis, it is observed that electric field is dominant on the outside of the proposed sensor (sensor 1), as seen in Fig. 2(a). Thus, it is useful for permittivity applications where the biosignal interacts directly with the sensor. The main field is used to form an effective capacitance with the biological tissues, where the outer capacitance $C_p$ represents...
the ability of biological tissues to store energy received from an external electric field. A change in the permittivity of the biosignal $\Delta \epsilon_{\text{tissue}}$ causes the outside capacitance to change $\Delta C_p$, $C_p \approx (\epsilon_{\text{tissue}})C_p^{\text{air}}$.

Using Kirchhoff’s voltage law (KVL), the input impedance $Z_{in}$ of the sensor 1 is derived from the external reader (as seen in Fig. 4(a)), which is given as:

$$Z_{in}(f) = \frac{V}{I_r} = j \left(\omega L_r - \frac{1}{\omega C_r} \right) - \left(\frac{\omega^2 L_s^2}{j(\omega L_s - \frac{1}{\omega C_s})}\right) + \left(-\omega L_{M_1}(\omega L_s - \frac{1}{\omega C_s}) - \omega^2 L_{M_{12}} L_{M_2}\right)\frac{\omega^2 L_{M_{12}}^2 - (\omega L_s - \frac{1}{\omega C_s})^2}{j(\omega L_s - \frac{1}{\omega C_s}) - j \omega L_{M_1}}$$

When the external reader is coupled to the passive sensor, then $f \approx f_s$, where the second, third, and fourth terms on the right-hand side in (8) causes a peak rise in the real part of the input impedance. Also, the reflection coefficient is represented as:

$$S_{11} = 20 \log |\frac{Z_{in} - Z_0}{Z_{in} + Z_0}|$$
where \( Z_0, L_r, \) and \( C_r \) are the characteristic impedance, inductance, and capacitance of the external reader, respectively \([16]\). \( L_{M1}, L_{M2}, \) and \( L_{M12} \) are the mutual inductance between the external reader and the top inductor, the external reader and the bottom inductor, and the top and bottom inductors of the sensor, respectively.

\( C_s \) represents the equivalent capacitance of the passive sensor, which in turn represents \( C_0 \), fringe capacitance, and the capacitance \( C_p \) of the outer surface of the inductor plate. From equations (5) and (6), we see that the electric field of the outer surface of the inductors plates is dominant, creating an effective capacitance, \( C_s \) is dominant. From equation (7), we see that the capacitance between the inductors plates is negligible because the electric field between the sensor is quite weak. The sensor resonates when the reactance is zero \((\text{Im}(Z_s) = 0)\). Fig. 5(c) shows the sensor circuit analysis whose resonant frequency can be expressed as:

\[
fs \approx \frac{1}{2\pi \sqrt{(L_s + C_s)}} \tag{10}
\]

In loose coupling between the external reader coil and the proposed passive sensor in the near field, the frequency splitting can not be observed. Thus, the resonance frequency can be further expressed as:

\[
fs \approx \frac{1}{2\pi \sqrt{(L_s + C_s)}} \tag{11}
\]

Considering the direction of the electric field of the inductor plate is outside the sensor towards infinity, \( C_p \) will not affect the distributed capacitance, and hence, \( fs = f_{sub} = f_1 = f_2 \). Further, by introducing the biological tissue, \( C_p \) is effective. In turn, \( C_s \) is effective, and the resonance frequency of the proposed sensor giving as:

\[
fs(\epsilon_t) \approx \frac{1}{2\pi \sqrt{(L_s + C_s(\epsilon_t))}}. \tag{12}
\]

**B. Two-symmetric inductor with opposite current direction**

Herein, repositioning one of the inductor of the proposed passive sensor is studied (sensor 2), as shown in Fig. 2(b). The turn direction of the top and bottom inductors is counter-clockwise and clockwise, respectively. Therefore, the current direction of the inductors is opposite to each other, which creates a high potential difference and an effective capacitance \( C_{in}^{eff} \) lies inside the inductors. Thus, the inductors store charges in between them and are electrically separated. The circuit analysis of sensor 2 is shown in Fig. 4(d) where the position of the dots on the terminal of each inductor is different, that is, at the opposite ends of the turn direction which indicates that the top and bottom inductors turns in opposite directions. As a result, the current leaving the dotted terminal of top inductor is 180\(^\circ\) out-of-phase with the current leaving the dotted terminal of the bottom inductor, indicating that the polarities of the voltages at the dotted terminals are also out-of-phase.

Therefore, when the voltage is positive at the dotted terminal of the bottom inductor, the voltage across the top inductor will be negative. This induces a high capacitance between the two inductors, where the net electric field on the inner surface of the inductors plates at \( RII \), as shown in Fig. 4(e), is given as:

\[
\oint \vec{E}_in \cdot da = E_{in}^{net}a \tag{13}
\]

\[
E_{in}^{net} = E_{in}^1 + E_{in}^2 = \frac{Q_1/2}{ \epsilon a} + \frac{Q_2/2}{ \epsilon a} = \frac{Q}{ \epsilon} \tag{14}
\]

Fig. 5(b) shows the electric field on the inner surface of the inductors plates. The electric potential is given as:

\[
V_{in} = E \int_a^{b} dL = E_s = \frac{Q_s}{e} \tag{15}
\]

The capacitance between the two inductors plates at \( RII \) can be expressed as:

\[
C_{in}^{eff} \approx \frac{Q}{ V} \approx \frac{eA}{s} \tag{16}
\]

Further, the overall inner capacitance between the inductors is given as:

\[
C_{in}^{eff} \approx \frac{eA}{s} \tag{17}
\]

where A represents the surface area of spiral metal trace and plate. Changes in the electrical separation \( s \) at any position of the inductor by an outside pressure or force lead to a linear change in the capacitance corresponding to the self-resonance frequency of sensor 2. Further, as observed in Fig. 4(d), the input impedance \( Z_{in} \) is given as:

\[
\begin{bmatrix}
V \\
0 \\
0
\end{bmatrix} = j
\begin{bmatrix}
\omega L_r - \frac{1}{\omega C_{in}^{eff}} & \omega M_1 & \omega M_2 \\
\omega M_1 & \omega L_s - \frac{1}{\omega C_{in}^{eff}} & \omega M_1 + \frac{1}{\omega C_{in}^{eff}} \\
-\omega M_2 & \omega M_1 + \frac{1}{\omega C_{in}^{eff}} & \omega L_2 - \frac{1}{\omega C_{in}^{eff}} \\
\end{bmatrix}
\begin{bmatrix}
I_r \\
I_1 \\
I_2
\end{bmatrix} \tag{19}
\]
The electric field between the plates at RII, as shown in Fig. 4(h) is given as:

\[ F_{\text{net}}^{\text{in}} = \frac{Q_1/2}{\varepsilon_0 a} \]  

The electric field outside the sensor at RII is given as:

\[ F_{\text{net}}^{\text{out}} = \frac{Q_1/2}{\varepsilon_0 a} \]  

As seen in Fig. 4(h) and 5(c), the electric field is dominant partly outside and partly inside the sensor. Therefore, we could say that the induced capacitance \( C_{\text{in}} \) in the sensor was due to the electric field inside the sensor, which could be suitable for sensing pressure or permittivity. However, reduced sensitivity is expected because there is no specific region where the electric field is dominant. From the circuit analysis shown in Fig. 4(d), the input impedance can be expressed as:

\[
\begin{bmatrix}
V \\
0
\end{bmatrix} = j \begin{bmatrix}
\omega L_r - \frac{1}{\omega C_r} & \frac{1}{\omega C_s - \frac{1}{\omega C_{\text{in}}}} & -\omega M_1 \\
\omega L_s - \frac{1}{\omega C_r} & \frac{1}{\omega C_s - \frac{1}{\omega C_{\text{in}}}} & -\omega M_1 \\
\end{bmatrix} \begin{bmatrix}
I_r \\
I_s \\
I_l
\end{bmatrix}
\]  

\[
Z_{\text{in}}(f) = j(\omega L_r - \frac{1}{\omega C_r}) - \left(\frac{\omega^2 M_2^2}{j(\omega L_s - \frac{1}{\omega C_r} - \frac{1}{\omega C_{\text{in}}})}\right) + \left(\frac{\omega M_1(\omega L_s - \frac{1}{\omega C_r} - \frac{1}{\omega C_{\text{in}}}) + \omega M_2(\omega M_1 + \frac{1}{\omega C_{\text{in}}})}{-(\omega L_s - \frac{1}{\omega C_r} - \frac{1}{\omega C_{\text{in}}})^2 + (\omega M_1 + \frac{1}{\omega C_{\text{in}}})^2}\right) \times \left(\frac{j(\omega L_s - \frac{1}{\omega C_s} - \frac{1}{\omega C_{\text{in}}}) + j\omega M_1}{j(\omega L_s - \frac{1}{\omega C_s} - \frac{1}{\omega C_{\text{in}}}) + j\omega M_1}\right).
\]  

Similarly, from the sensor circuit analysis shown in Fig. 4(i), the resonance frequency of sensor 2 is given as:

\[
f_s \approx \frac{1}{2\pi \sqrt{L_s + L_{M2}(C_s + C_{\text{in}})^2}} \approx \frac{1}{2\pi \sqrt{L_s(C_s + C_{\text{in}})}}.
\]  

C. Inductor on the top of the plate

This section discusses another modification that was made to the proposed passive sensor, which is replacing one of the resonators with a plate (sensor 3), shown in Fig. 2(c). The external reader is inductively coupled to the inductor with current flowing through the inductor, as shown in Fig. 4(g). The electric field between the plates at RII, as shown in Fig. 4(h) is given as:

\[
E_{\text{net}}^{\text{in}} = \frac{Q_1/2}{\varepsilon_0 a} \]  

Thus, the resonant frequency is expressed as:

\[
f_s \approx \frac{1}{2\pi \sqrt{L_L(C_s + C_{\text{in}})}}.
\]  

Figure 6 shows the self-resonance frequency of the three passive sensors with the presence of the FR4 slab between the plates. A large frequency shift is observed in sensor 2 as compared to sensors 1 and 3 due to the intensity of the uniform electric field between the inductors, which produces \( C_{\text{eff}} \) as given in equation (17), thus increasing the equivalent capacitance of sensor 2. However, there is no frequency shift in the proposed passive sensor (sensor 1) because there is no effective capacitance between the inductors due to no electric field (see Fig. 5(a)). Additionally, there is no energy stored in the substrate, which is an advantage in monitoring the permittivity change of the biosignals. Further, the effect of change in the separation \( s \) between plates is studied. Shown in Fig. 7(a), no frequency shift was observed for the proposed passive sensor (sensor 1) when separation \( s \) is reduced. However, the resonance frequency of sensor 2 decreases with decrease in the separation between the two plate, as seen in Fig. 7(b), and proven in equations (16) and (22). Although the resonance frequency of sensor 3 decreases with change in the separation due to \( C_{\text{in}} \), it has lower sensitivity than sensor 2, as shown in Fig. 7(c). Furthermore, surrounding the sensors with lossy material such as water is studied. As shown in Fig. 8(a), the proposed passive sensor (sensor 1) shows the maximum resonance frequency shift due to the dominant strength of the uniform electric field outside the sensor, which creates an effective capacitance with lossy material (water). As a result, \( C_s \) is effective. As shown in Fig. 8(b), a slight resonance shift is observed in sensor 2 mainly due to the fringe field of the inductors lines. There is a noticeable resonance shift in sensor 3, as shown in Fig. 8(c).

### III. Measurements and Results

A. Pressure monitoring (wound monitoring)

Different internal pressures always correlate to change in the separation distance of the conductive plates of the capacitor, regardless of using wired [47], [48] or wireless sensors [49], [50]. Pressure sensing can also be used for wound monitoring. The passive sensor is placed on the surface of the wound...
Thus, we studied the inductor on the top of the plate (sensor studied by using an inductor on the top of the plate [29]. Conductive plates give rise to any changes in the separation distance of the sensor’s parallel plates on its resonance frequency. Wound monitoring has been previously caused by the pressure. Thus, we studied the effect of change in separation distance on the resonant frequency changes when pressure is introduced. An effective capacitance between the conductive plates varies inversely with their separation. Thus, the frequency changes when capacitance between the conductive plates varies inversely with the separation, which is introduced. Effective capacitance between the conductive plates gives rise to any changes in the separation caused by the pressure. Thus, we studied the effect of change in separation distance of the sensor’s parallel plates on its resonance frequency. Wound monitoring has been previously studied by using an inductor on the top of the plate [29]. Thus, we studied the inductor on the top of the plate (sensor

Fig. 7. HFSS-simulated resonance of the passive sensor by changing separation $s$ between the plates by changing the dielectric material thickness. (a) Sensor 1 (b) Sensor 2 (c) Sensor 3

Fig. 8. HFSS-simulated resonance of the passive sensor with a thickness of 0.2 mm by changing the surrounding medium. (a) Sensor 1 (b) Sensor 2 (c) Sensor 3

Fig. 9. Top view of a passive sensor

Fig. 10. Measured and simulated the values of the resonance frequency of passive sensor by changing the separation between the plates. (a) Sensor 1 (b) Sensor 2 (c) Sensor 3

Fig. 11. External reader noise with sensor resonance frequency through phantom

Fig. 12. (a) Measured $S_{11}$ response of sensor 2 with different thicknesses inside the muscle tissue. (b) Estimated linear values of resonant frequencies for different thicknesses.

Fig. 13. (a) Measured sensor resonance for different glucose levels (sensor 1). Proposed sensor immersed in different glycaemic fluid levels. (b) Estimated linear values of resonant frequencies for different glucose levels.

and monitors mechanical pressure on the wound position [51]. In a wireless, battery-free resonator-based biosensor, the resonant frequency varies as the inverse square root of the capacitance between the conductive plates varies inversely with their separation. Thus, the frequency changes when pressure is introduced. An effective capacitance between the conductive plates gives rise to any changes in the separation caused by the pressure. Thus, we studied the effect of change in separation distance of the sensor’s parallel plates on its resonance frequency. Wound monitoring has been previously studied by using an inductor on the top of the plate [29]. Thus, we studied the inductor on the top of the plate (sensor
human tissues, especially for high frequencies and confirming monitoring. To determine the advantages of the sensor 2, Table III shows each sensor indicates that the results can be reproduced. To as proven in equations (14), (16), and (22). A slight increase capacitor, where the electric field is dominate inside sensor 2 thicknesses (i.e., top view of the passive sensor in Fig. 9). In comparison to sensors 1 and 2 in the same environment conditions (i.e., external reader, substrate). To confirm which the passive sensor is suitable for monitoring the healing of the wound, three passive sensors with different separations s between the plates were designed and fabricated. The separation s between sensor plates varied from 0.2-0.5 mm with substrate thicknesses ranging from 0.2-0.5 mm substrate thicknesses (i.e., top view of the passive sensor in Fig. 9).

$S_{11}$ were measured and the resonant frequencies of the sensors were recorded wirelessly from the external reader attached to the vector network analyzer (VNA) via a coaxial cable. Responses from $S_{11}$ were received in the frequency range of the resonance of the sensor, which was recorded. Each thickness was measured thrice to ensure consistency in the results. From the results, it is confirmed there is no correlation between the separation distance of the plates, and the change in resonance frequency of the proposed passive sensor (sensor 1), as shown in Fig. 10(a), and proven in equation (4). Furthermore, the simulation results are in good agreement with measured results. However, in sensor 2 and sensor 3, the resonance frequency decreases with decrease in s, as shown in Fig. 10(b) and 10(c). The shift in the frequency f is directly proportional to the decrease in the separation s of the passive sensor, by an excellent correlation coefficient, $R^2 = 0.9989$ and $R^2 = 0.99779$, for sensors 2 and 3, respectively. Moreover, sensor 3 has a sensitivity of 99 MHz/mm, and sensor 2 has a sensitivity of $2.5 \times 10^3$ MHz/mm, which is 25 times higher than that of sensor 3. Thus, sensor 2 is most reliable for monitoring wounds as well as various internal pressures. The positive relationship between resonance shift and s with high sensitivity is caused by an effective inner capacitor, where the electric field is dominate inside sensor 2 as proven in equations (14), (16), and (22). A slight increase in the error bar on conducting three different experiments of each sensor indicates that the results can be reproduced. To determine the advantages of the sensor 2, Table III shows a comparison between the three passive sensors for wound monitoring.

Because of the absorption issue of the EM field on the human tissues, especially for high frequencies and confirming the possibility for sensor 2 for implanted applications for various pressure application, a new measurement technique using sensor 2 was prepared, wherein the sensor 2 was injected inside the phantom muscle tissue at a depth of 2 mm. The sensor was covered by an insulator material to prevent it from short-circuits when it interacting with the tissue. The external reader was placed at a distance of 1 mm from the tissue and 3 mm from the sensor. A tissue sample was then prepared, as given in [53]. In wireless measurements, the external reader introduces noise, as shown in Fig. 11. Thus, the noise was subtracted from the external reader signal by using the derived noise rejection (NR) equation, which is defined as:

$$NR(f)[dB] = |(S_{11}(f))_{rws} - (S_{11}(f))_{ro}|$$

Table III. Comparison between the three passive sensors for wound sensing

| Passive sensor | Description | Linear equation $f$ in (MHz/mm), s in (mm) | Sensitivity (MHz/mm) |
|---------------|-------------|------------------------------------------|---------------------|
| Sensor 1      | Two resonators identically charged | - | - |
| Sensor 2      | Two resonators oppositely charged  | $f = 2.107 + 2.5951s$ | $2.5 \times 10^3$ |
| Sensor 3      | Inductor on top plate | $f = 4.07 + 0.0997s$ | 99 |

Glucose concentration level plays an important role in diabetes, and hence, change in glucose levels was also studied. A commercial electrochemical technique known as finger-pricking devices, is commonly used to detect the glucose levels in the blood. However, in this method, the patient’s finger is
pricked many times in the day, and hence, it can be painful for the patient. Therefore, we require new technologies that are non-invasive or can be used as long-term implantable devices, such as the proposed miniaturized passive sensor that is suitable for non-invasive and minimally invasive applications, and interacts with the interstitial fluid to monitor the glucose levels. The outside capacitance interacts with the glycaemic fluid and creates an effective capacitance. In previous studies, aqueous glucose solutions were used for the initial experiments to monitor different glucose concentrations using microwave sensors because the concentration of the glucose is the most dominant as compared to the other components of the blood. Further, the glucose level in the interstitial fluid responses to changes in blood glucose [52], [55]–[58]. Thus, the glucose concentrations in aqueous solutions were prepared within the range of 0-500 mg/dL. 500 mg/dL of the stock solution was first prepared by mixing glucose powder and deionized water whereas the other concentrations were made using deionized water and stock solution.

The proposed passive sensor was immersed in a glycaemic fluid. Using a micropipette, 60 µL glycaemic fluid was dropped into the container. The glucose concentration was wirelessly recorded using an external reader from a distance of 3 mm from the passive sensor. The expansion of the errors bar represent the errors noted in the three different measurements of glucose levels. Moreover, the equation comprises a statistical $R^2 = 0.9576$, with a sensitivity of 500 kHz/(mg/dL) considering different glucose levels in the aqueous solution.

The proposed passive sensor (sensor 1) was further studied for use in wearable applications in the future. The passive sensor was covered with a thick plastic tape to avoid direct contact between the solution and sensor. The distance between the sensor and external reader was approximately 3 mm. The glycaemic fluid with a volume of 30 µL was dropped on the top of the sensor. Further, additional different glycaemic solutions were dropped on the top of the wrapping that covered the passive sensor. This wrapping served as a model for the skin barrier. The field from one side of the sensor could penetrate the thick tape to interact with the glycaemic fluid. The experiment was repeated three times. An increase in frequency was observed, proving that as the concentration of glucose increases, the frequency increases with a sensitivity of 46 kHz/(mg/dL), as shown in Fig. [14](a) and (b). Thus, we were able to demonstrate that the glucose level of the fluid can be observed by invasive or non-invasive techniques using the proposed passive sensor.

Table IV shows a comparison between different types of resonators. The proposed sensor was further scaled down to unprecedented dimensions of 2 mm × 2 mm × 0.2 mm to achieve the title of smallest biosensor reported for glucose monitoring, this requiring the smallest amount of tissue sample to detect the glucose levels. Further, introducing a wireless, battery-free resonator-based glucose monitoring provides extra advantages than other recently reported works [18]–[22]. Further, it has better sensitivity than the previous wired sensors, [12], [21]. The proposed sensor has higher sensitivity at most

| [Ref.], Year | Resonator type | Sensing method | Structure volume (mm³) | Sample volume (µL) | Concentration (mg/dL) | Type of application for diabetic | Sensitivity (MHz/(mg/dL)) |
|-------------|----------------|----------------|------------------------|--------------------|------------------------|-------------------------------|--------------------------|
| [12], 2021  | Localized spoof surface plasmon | Wired | $14 \times 5.03 \times 1.52$ (Partial volume) | 100 | 0-50000 | Outside the glycemic range | $1.2771 \times 10^{-6}$ |
| [18], 2020  | Split ring | Wired | $9.6 \times 6 \times 0.5$ | - | 50-400 | Non-invasive | 3.5 |
| [19], 2020  | Split ring | Wired | $66 \times 20 \times 0.8$ | 600 | 70-110 | Non-invasive | 0.94 |
| [20], 2020  | Split ring | Wired | $66 \times 30 \times 0.8$ | 100-600 | 80-120 | Non-invasive | - |
| [21], 2020  | Complementary electric LC | Wired | $80 \times 100 \times 3.175$ | 95 | 100-500 | Non-invasive | 0.0185 |
| [22], 2021  | Coplanar waveguide | Wired | $30 \times 18 \times 0.508$ | - | 100-500 | Non-invasive | 3.53-3.58 |
| [53], 2019  | Inductor on top plate | Wireless | $4 \times 4 \times 1$ | - | 0-500 | Invasive | 0.015 |
| [54], 2020  | Cavity | Wireless | $4 \times 4 \times 2$ | 4000 | 75-250 | Invasive | 0.032 |
| Proposed sensor | Parallel resonators | Wireless | $2 \times 2 \times 0.2$ | 30-60 | 0-500 | Non-invasive and invasive | 0.046-0.5 |
30 times and 15 times more than previous wireless sensors, \cite{53}, \cite{54}, respectively. It is compact in size, light weight, which gives it a possibility to be applicable in minimally invasive or non-invasive techniques, as compared to other resonators in Table IV. The proposed passive sensor is a promising candidate for implantable or wearable applications that monitor glucose concentrations in the interstitial fluid. Wireless, battery-free, and sensor-based glucose monitoring is a fundamental step for the research underway, and there is scope for commercialization and presenting more suitable and accurate glucose monitoring. For example, we can calibrate the measurement of the proposed passive sensor for non-invasive technique and invasive technique by developing a library of the measured frequency response curves under a large variety of glucose concentrations when the passive sensor is implanted underneath the skin or attached to the skin with help of volunteers.

IV. CONCLUSION & FUTURE SCOPE

By simply changing the position of the inductor of the proposed passive sensor based on two parallel resonators, we can change the coupling between the passive components of the sensor entirely, as well as the region of the dominance of the electric field. Therefore, by knowing the region of the dominance of the electric field the passive sensor is no longer limited to pressure sensing, which proves that using two parallel resonators is more effective than using an inductor on the top of the plate, where the electric field is not dominant. Three miniaturized passive sensors based on the parallel-plate technique were fabricated to wirelessly monitor the wide-range biosignals, where the effective capacitive of each passive sensor was studied. The two symmetric inductors of the passive LC tank sensor with opposite direction current flow (sensor 2) created a high capacitance between the inductors, which was suitable for wound monitoring. Whereas same direction current flow (sensor 1) created a capacitance outside the inductors, making the sensor sensitive to the permittivity change of biological tissues, such as glycaemic fluid.

The miniaturized passive sensors based on parallel resonators can be used for minimally invasive and non-invasive techniques, and can be easily injected under the skin using a rigid substrate for permittivity monitoring, i.e., sensor 1, or a flexible substrate for pressure monitoring, i.e., sensor 2. It can also be used as a conformal structure wrapped around a specific region of the body without introducing other elements to improve its capacitance. The passive sensors based on parallel resonators are promising for continuous, convenient, and faithful biosignals monitoring and an excellent alternative for all wired active resonator-based biosensors in wearable and implantable applications. Further reader development and introducing the concept of PT-symmetric and EPs to the system will improve the sensing capability of passive sensors and open the door for a new generation of biosensors.

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