RESEARCH ARTICLE

A low complexity RF based sensor array for lung disease detection using inkjet printing

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Abstract

This article presents the design of a low complexity, low power, small-sized, wearable, inkjet printed, cheap, and noninvasive radio frequency (RF) based sensor array for pulmonary edema and emphysema condition monitoring inside the patient chest. The RF sensor consists of 38 electrodes and 37 ports. The size of the sensor is 4 cm × 89.4 cm to cover the chest of an average adult. The sensor is optimized to operate at 60 MHz. The scattering coefficients S_{i1} are measured at each passive port and then the Least Squares (LS) method is used to form an equation for average dielectric constant estimation. The dielectric constant estimation method is used to detect the presence of water/air in human and porcine lungs. The average measured dielectric constants of normal human lung tissue, edema, and emphysema infected lungs are estimated with errors of 3.54%, 4.83%, and 4%, respectively. The porcine lung tissue-mimicking phantom with proper electrical properties is formed using a water and salt (NaCl) mixture. To detect the different stages of pulmonary edema, 200 mL water balls are inserted in the inner layer of the chest model. The measured errors were 2.68%, 0.87%, 2.18%, and 2.8% for normal porcine lung, adding 6 water balls, adding 12 water balls, and adding 18 water balls, respectively.

KEYWORDS

dielectric constant, inkjet printing, least square method, pulmonary edema, pulmonary emphysema, RF sensor

1 | INTRODUCTION

The traditional methods of noninvasive imaging in the medical field are the magnetic resonance imaging (MRI), computed tomography (CT), and X-rays. These methods are expensive and require heavy equipment. The frequent use of CT and X-rays methods will affect the patients' health due to radiation exposure. Unattended and uninterrupted monitoring of a patient health is an essential requirement in some medical devices. Some commercially available devices can monitor respiration rate, temperature and heart rate.^{1,2} Using the Doppler Effect, printed antennas for body wearable applications were used for remote detection of respiration and heartbeat in Ref. 1. Other wearable health monitoring devices are also available, but they cannot monitor deep tissue changes and many of them are very expensive. A low-

cost, wearable sensor capable of continuously monitoring a patient condition (ie, deep tissue conditions) is of great importance in the medical field.

The traditional methods for noninvasive imaging of deep tissues within the body have many disadvantages such as unsuitability for continuous time monitoring, need for regular hospital visits, expensive, difficult to carry due to large and heavy equipment, and side effects due to radiation exposure if frequently used.² Wearable and continuous health monitoring sensors are discussed in Refs. 2,3. These are limited for monitoring breathing, heart rate and skin temperature only. These sensors have advancements in data collection but they lack deep tissue monitoring.

Noninvasive imaging techniques include electrical impedance tomography (EIT), ultrasound impedance tomography and microwave imaging.⁴ By measuring the voltages

and currents on the periphery of a body, EIT reconstructs the spatial conductivity and permittivity within the body.^{4–7} Abnormalities in the human body can be detected by recording the changes in the measured voltages or potential differences in the body density. Its accuracy is low due to tissue diversity and its inability in reconstructing complex shapes.

Microwave imaging is used to recover exact electrical properties (permittivity and conductivity) in the frequency range of 300 MHz to 30 GHz. In this technique, an array of wide band antennas is placed around the body to collect the back scattered and bi-static data. The main issues when using this technique are the high frequency of operation and small penetration depth. This technique is applicable for near fields only. It is used for breast cancer detection and treatment response monitoring.⁸ The portability and wearability are not possible due to the complex circuitry needed for sending and receiving wide-band signals. The accuracy is affected by the lossy characteristics of the biological tissues at microwave frequencies. Ultrasound techniques are based on the acoustic properties of the tissue and rely on the transmission, diffraction, and reflection of acoustic wave within it. Its sound frequency range is low (1–20 MHz) to ensure deep tissue penetration.⁹ The echoes are separated by the time delay of reception and the received signal strength. This technique is mostly suitable for soft tissues and not for bones (ie, abdominal, cardiac, urological, etc). The ultrasound device can be portable but not wearable. Its accuracy is limited. In short, EIT and ultrasound techniques give a contrast image rather than actual values of the electrical properties while the microwave imaging method has high frequency of operation hence small penetration depth within the tissue.

Microwave sensors were presented in Refs. 10–12 with an operating frequency of 915 MHz and wearable health monitoring sensors were proposed in Refs. 13–15 and operated at 40 MHz. They had a planar geometry, high error, and large size when estimating the permittivity of a small area. Some wearable wrapped around a circular cylindrical chest model sensors were also proposed in Refs. 15,16, but the actual human chest is elliptical. Also, sub-grouping of the sensors makes the post processing of the data difficult.

Our proposed on-body monitoring system combines an RF wearable sensor with deep tissue monitoring via the changes in the electrical properties of the tissue along with its post processing algorithm. This is beneficial for patients at risk of having pulmonary edema (due to fluid accumulation in lungs) and emphysema conditions (due to air accumulation in lungs). The sensor is to be used to continuously monitor the patient condition with minimum exposure to electromagnetic (EM) radiation. The sensor will keep track of the patient condition via determining the average dielectric constant value of his/her chest. The sensor operates at 60 MHz, has small dimensions, is low cost, and of low complexity. The substrate used is a thin polyimide sheet which is flexible and can be integrated within wearable

cloths (T-shirts). An inkjet printer is used for the sensor fabrication. The input power and operating frequency are low, thus reducing health risks. Moreover, the RF sensor is noninvasive and nonintrusive and its size is $4 \times 89.4 \text{ cm}^2$ covering the complete chest. The chest model used in the measurements is elliptical and close to the average adult chest size. The average error of all measured human and porcine lung cases was less than 5% making it the most accurate using a simple algorithm. The sensor can monitor the inner tissue conditions of the human chest as deep as 11.64 cm. The achieved dynamic range for low to high ϵ_r variation is equal to 20 dB which is much higher than previous works. It can detect the presence of water in the whole area of the human upper chest.

The rest of the article is organized as follows: a brief literature survey is shown in Section 2 with a comparison table with all related works. The details of the RF sensor design are given in Section 3. Section 4 presents the dielectric constant estimation method. The measured electrical properties of different liquid phantom combinations using a dielectric assessment kit (DAK) are presented in Section 5. The experimental results for the human and porcine lung properties are gathered in Section 6. Finally, conclusions are given in Section 7.

2 | LITERATURE SURVEY

A complete literature survey was conducted and the features of various biomedical sensors are tabulated in Table 1. EIT, microwave imaging and ultrasound impedance tomography based sensors suffer from large size, high error, and portability issues.^{4–9} Several on body sensors operating at 915 MHz were proposed in Refs. 10,11 while some showed textile based integration.¹² These sensors had health risk due to high operating frequency, wearability issues, and can be applicable for monitoring a small area only.

Several low frequency based sensors were proposed operating around 40 MHz. Some had a planar geometry, large size, errors more than 8%, and small covered areas,^{13,14} while others were wrapped around a circular chest model geometry with high complexity processing method (ie, grouping of estimates) and high error percentages.^{15,16} Some textile based sensors were also proposed in Refs. 17,18, they suffered from large size, high cost, and small area monitoring capabilities due to their planar geometry. A wearable sensor was proposed in Ref. 19 for measuring the respiratory rhythm of young children to detect respiratory disorder such as choking and asthma. The sensor was based on the variation of resistance of a knitted electrically conductive fabric, integrated into baby underwear. The resistance studies of many samples were performed and an optimum type of stitch used for knitting was selected. The drawback of such a method is the selection of the optimum type of stitch because it strongly impacts the properties of the respiratory rhythm.

TABLE 1 Comparison of biomedical sensors available in the literature

Reference	Category	Application	Frequency (MHz)	Size (cm)	Complexity	Error (%)	Cost	Limits
3	Fabric sensors using advance signal processing techniques	Cardiovascular diseases	–	Large	High	–	Very high	Limited for monitoring breathing, heart rate and skin temperature, advance in data collection but lack in deep tissue monitoring
4–7	Electrical impedance tomography (EIT)	Real time imaging of ventilation, detection of collapsed lung	1 to 20 000	Very large	High	High	Very high	Limited to bed side application due to complex circuitry for generation of variable current patterns
8	Microwave imaging	Breast cancer detection and treatment response monitoring	300 to 30 000	Small	High	High	Low	Small depth of penetration, portability and wearability is not possible due to complex circuitry
9	Ultrasound impedance tomography	Abdominal, cardiac, urological, fetal, pediatric and fluid detection measurement and analysis	(1–20)	Very large	High	High	Very high	Portable but not wearable
10	Microwave reflection and transmission measurements	Pulmonary diagnosis and monitoring	915	Very large	Very high	–	High	Health risks due to high frequency operation, not wearable
11	Microwave sensor	Breathing, heart rate and changes in lungs water contents	915	–	Very high	–	Low	Health risks due to high frequency operation, not wearable, short time Fourier transform based digital signal processing makes it more complex
12	Textile EM coupler	Lung water contents monitoring	915	4 × 24	Low	–	High	Monitor the permittivity of a small area only
13	RF sensor	Pulmonary edema monitoring	40	10 × 16	Low	11	Low	Monitor the permittivity of a small area only
14	RF sensor	Lungs dielectric properties	40	10 × 16	Low	8	Low	Monitor the permittivity of a small area only
15	RF sensor with pixel by pixel imaging	Cardiovascular diseases and tumor growth monitoring	40	5.3 × 113	High	6	Low	Experimental verification was not done
16	Flexible RF sensor	Monitoring deep tissues electrical properties	40	5 × 83	High	13.5	Low	Subgrouping of the sensor makes post processing of the data difficult
17	Textile sensor	Wearable applications	40	–	Low	–	High	Monitor the permittivity of a small area only
18	Textile sensor	Wearable applications	40	10	Low	–	High	Monitor the permittivity of a small area only
Proposed	Flexible RF sensor	Pulmonary edema and emphysema monitoring	60	4 × 89.4	Low	4.83	Low	–

In addition, washing or folding the sensor may produce the wrong results/misdetected. A multi-slot coaxial antenna operating at 2.45 GHz was proposed in Ref. 20 to optimize the scattering parameters by location and size of the slots in antenna structure and limit the antenna input power to enable appropriate adjustment of the temperature of the target tissues for cancer treatment. It suffers from causing health effects due to its high frequency of operation.

Previously, we proposed an RF sensor array for estimation of multiphase oil flow in pipelines.²¹ In this current work, we propose an RF sensor array for lung disease detection with a totally different geometry, fabrication process,

and experimental validation method with a lower operating frequency. The proposed sensor also has a higher dynamic range compared with Ref. 21 and; thus, has much higher accuracy (error <5%).

Compared with previous studies listed in Table 1, our proposed biomedical RF sensor has many beneficial features. It is nonintrusive, noninvasive, and smaller than others applicators. Moreover, it is inexpensive and simple to implement. The sensor is wrapped on an elliptical chest model which represents an actual scenario unlike all others that use cylinders. The sensor is robust and can be integrated within clothes/T-shirts. It can estimate different lung diseases/

conditions with the lowest measured error percentages compared with all mentioned works. It can monitor the inner tissues as deep as 11.64 cm from its surface as compared with 8 cm in Ref. 13. A single sensor array covers the whole area of an average adult chest, no sensor sub-grouping is required thus simpler data processing is required. The achieved dynamic range between low and high values of dielectric constant ϵ_r is 20 dB compared with 12 dB in Ref. 21. The real time ϵ_r -measurements are used for the experiments using dielectric assessment kit (DAK) setup. Both homogeneous and nonhomogeneous experimental cases are discussed unlike most related works.

It should be noted that proposed RF sensor will not be able to specify the water content location but rather track the changes in its volume. It will provide continuous patient monitoring that cannot be achieved using other methods due to the harm caused by the radiation exposure.

3 | RF SENSOR DESIGN

The RF sensor consists of 38 electrodes and 37 ports. The first port is excited with 60 MHz RF signal and the remaining 37 ports were terminated with 50-Ohm coaxial probes except the port where the measurement is taken.¹⁴ The passive receiving ports cannot act as exciters due to their small

sizes. Biological issues show significant variation of ϵ_r below 100 MHz¹⁴ so the operating frequency was optimized to 60 MHz. This frequency was chosen based on the maximum achieved dynamic range for the material properties variation considered. A flexible substrate (polyimide sheet) is used with a thickness of 50 μm . The sensor was designed and optimized using a commercially available ANSYS HFSS software. The simulations were conducted for the frequency range 1-100 MHz with a resonant frequency of 60 MHz. The width of first two electrodes is 4 cm and other electrodes have a width equal to 2 cm. The substrate dimensions were $4.4 \times 89.4 \times 0.005 \text{ cm}^3$. The sensor design is based on the idea introduced in our previous work²¹ and other works^{13,14} with a totally different geometry. Several designs were investigated, and the proposed design gave the best dynamic range response and smallest size. The first two electrodes were designed to operate at 60 MHz under the best possible matching conditions to perform as the signal exciter. The remaining 36 electrodes receive the radiated fringing fields. The gap between the electrodes was optimized to 0.22 cm. The proposed design with detailed dimensions is shown in Figure 1A,B.

The sensor was wrapped around an elliptical average adult human chest upper torso model with two ellipse axis given as 33.2 cm and 23.28 cm. The upper torso of the human body has a length of 18.29 cm.²² The perimeter of an

FIGURE 1 Proposed designed planar RF sensor: (A) complete view, (B) enlarged view for first 10 electrodes, (C) sensor wrapped around an averaged adult chest model, and (D) comparison between scattering parameters S_{11} for different values of electric permittivity ϵ_r

TABLE 2 Electrical properties of different materials used for designing of RF sensor

Material	Thickness (cm)	Relative permittivity	Loss tangent	Electric conductivity (S/m)	Reference
Polyimide sheet	0.005	3.4	0.01	–	24
FR-4 sheet	0.07874	4.8	0.017	–	25
Plexiglas base	1	2.5	0.08	–	26
Outer layer	1.5	40.6	2.33	0.38	14,15
Inner layer	31.62	1–120	2–3	0.3–0.5	14

average human chest is 89.4 cm (which is also the substrate width of the RF sensor) having an elliptical shape with major radius 16.6 cm and minor radius 11.64 cm.²³ The average thickness of the outer chest layer (skin, fat, muscles, and bones) is 1.5 cm.¹⁴ The wrap around design of the proposed RF sensor is shown in Figure 1C. The electrical properties of the substrate, FR-4 sheet, plexiglas base, outer layer, and inner layer of the chest model are summarized in Table 2. If outer layer properties change, the model needs to be re-simulated to extract new curves and update the ϵ_r -estimation equation. The dielectric constant of the inner layer was varied from 1 to 120. The scattering coefficients (S_{i1}) were obtained for different values of electrical permittivity (ϵ_r), namely 1, 30, 60, 90, and 120 inside the chest, where i is the port number. The comparison between S_{i1} curves of different ϵ_r values is shown in Figure 1D. The maximum achieved dynamic range between low and high ϵ_r values was 20 dB (which is the difference of largest and smallest scattering coefficient values at port 17) at port 17, while the achieved dynamic range in Refs. 14,16, using ϵ_r values from 1 to 180 was 22 dB for planar or semi planar sensors. The dynamic range is the range (in dB) associated with the most affected port by the inner layers (port 17 in our case) from the active one for the two extreme ϵ_r values expected, that is, 1 and 120, in current investigation. The dynamic range of S_{i1} should be large enough to accurately map the dielectric properties. With large dynamic range, we can detect small changes in ϵ_r more precisely which is an essential aspect of monitoring illness or recovery. There is no need for RF sensor sub-grouping as done in Ref. 16 which makes the post processing of the data difficult. By changing the ϵ_r values, a shift in S_{i1} curves was observed which will improve its accuracy for ϵ_r estimation using the Least Squares (LS) method.

3.1 | Specific absorption rate

Specific absorption rate (SAR) is a metric that evaluates the safety limits for different devices producing EM radiation inside the human tissue. SAR should be evaluated for wireless medical devices and all kinds of cell phones. The power absorbed by the tissue surrounding the antenna/sensor

should not exceed the standard limits. According to IEEE, the 10 g averaged SAR value should not exceed 2 W/kg. The Federal Communication Commission (FCC) limits 1 g average SAR value to 1.6 W/kg. According to IEEE and FCC recommendations, the whole body SAR value is limited to 0.08 W/Kg.^{27–29} The SAR-parameter of the designed RF sensor was calculated using ANSYS HFSS internal engine. SAR is defined as follows:³⁰

$$\text{SAR} = \int_{\text{Sample}} \frac{\sigma(r)|E(r)|^2}{\rho(r)} dr \quad (1)$$

where, $|E(r)|^2$ is square of the electric field strength, σ is the tissue sample r electrical conductivity, and ρ is the mass density of the tissue sample. We calculated the local SAR. The ANSYS HFSS software uses Equation (1) to calculate the SAR at each mesh point on an overlay plot. It then interpolates the values between the mesh points across the plot. The SAR values were calculated in watts per kg. The SAR was evaluated with an applied transmitting power of 1 W. The mass of tissue was kept 1 g and material density 1 g/cm³.³¹ The SAR values are greatly affected by the applied transmitting power level.³¹ In this work we considered that the outer layer electrical properties remained the same. The SAR was calculated by changing the inner layer electrical properties only with maximum conductivity value of 0.5 S/m to analyze the worst case scenario of SAR. The maximum value of SAR was always near the first port (exciting source) for all analyzed cases, namely for $\epsilon_r = (1, 30, 60, 90, 108, 120)$ and the obtained values were far away from the FCC and IEEE standards as shown in Table 3.

We used a 1 cm thick plexiglas sheet to form the base of the modeled human chest torso. The base is used to hold the inner layer and outer layer separation FR-4 sheets of 0.8 mm thickness. These FR-4 sheets were added in the simulation model and their effect was considered. The base was not considered in SAR calculations (note that the radiation is inside the ellipse and not toward the base). By adding the plexiglas base to the simulation mode, the SAR values are not expected to be affected as the sensor radiation is not toward it and its far away from the sensor center (vertical distance from center of the sensor to base is 9.15 cm).

TABLE 3 SAR values calculated using ANSYS HFSS software

Inner layer	$\epsilon_r = 1$	$\epsilon_r = 30$	$\epsilon_r = 60$	$\epsilon_r = 90$	$\epsilon_r = 108$	$\epsilon_r = 120$
SAR (W/kg)	0.035692	0.0097286	0.0088508	0.0071432	0.0077559	0.0054433

FIGURE 2 RF sensor printing steps: (A) substrate washing with isopropyl alcohol (IPA), (B) drying with nitrogen gas, (C) UV light curing, and (D) substrate fixing on inkjet printer

3.2 | Sensor fabrication using silver inkjet printing

The proposed RF sensor was fabricated using a silver inkjet printer at the IMPACT lab, King Abdullah University of Science and Technology (KAUST), Saudi Arabia. The sensor was printed using advanced nano products (ANP) silver ink. Since the available inkjet printer can print only A4 size sheets, the proposed sensor was divided into five parts. The first step was washing the substrate with isopropyl alcohol (IPA). The second step was the drying process of the polyimide substrate that was performed using nitrogen gas. In the third step, ultra-violet (UV) light curing for 1 min time period was performed for good silver layer printing on the polyimide. Increasing the UV light curing time will increase the silver ink spreading on the substrate. The forth step was fixing of the substrate on inkjet printer. All these steps are shown in Figure 2. Three ANP silver ink layers were printed to achieve good conductivity values. After each layer

printing, 10 min heating is applied at 130°C and after the last layer printing, heating time is increased to 15 min. The drop spacing of the ANP ink during printing was optimized to 30 μm . Decreasing the drop spacing results in ink spreading on the substrate.

The five fabricated parts of the sensor were joined using double sided tape. The final fabricated RF sensor prototype without coaxial probes is shown in Figure 3A, the flexibility of the RF sensor is in Figure 3B on a shirt flexible material, the scaled RF sensor with 30 cm scale is presented in Figure 3A,D shows the fabricated RF sensor with 50-Ohms coaxial probes attached to each port. The constructed model is used as a proof of concept model. In a commercial one, the probes can be replaced with smaller ICs that are capable of power measurements and may wirelessly transfer their values to a nearby processing unit.

3.3 | Human chest upper torso design

The human upper chest torso model was designed in the Fablab at King Fahd University of Petroleum and Minerals (KFUPM), Saudi Arabia, using a computer numerical control (CNC) machine. A plexiglas sheet of 1 cm thickness was used for the human chest torso base. Two elliptical grooves for the outer and inner layers were modeled with 0.5 cm thickness to fix the flexible 0.8 mm FR-4 sheets that were used to separate the inner and outer layer solutions from one another. The sheets separating inner and outer layers were made of FR-4 material and they were modeled in ANSYS HFSS software during the simulations to take their effect into the dynamic range and LS method calculations and to reflect the measurement setup. The height of the human upper chest torso was 18.29 cm. Super glue was used to join the FR-4 sheets and epoxy glue was used to fix the FR-4 sheets in the grooves. Figure 4A shows the plexiglas base design on the CNC machine with software commands running on the computer. Figure 4B-D shows the top, front, and side views of the designed human upper chest elliptic torso, respectively.

FIGURE 3 Fabricated RF sensor: (A) RF sensor without coaxial probes, (B) flexible RF sensor on a shirt, (C) scaled RF sensor compared with 30 cm ruler, and (D) RF sensor with coaxial probes

FIGURE 4 Human chest upper torso phantom design: (A) plexiglas base design, (B) top view, (C) front view, and (D) side view

4 | AVERAGE DIELECTRIC CONSTANT ESTIMATION

The dielectric constant can be estimated using a similar expression as below¹⁴ Equation (2),

$$\epsilon_r = \sum_{i=2}^{37} w_{i-1} S_{i1} \quad (2)$$

Following Equation (2)^{13,14} w_{i-1} are the weight coefficients, S_{i1} are the scattering coefficients, and i is the port number. S_{i1} distributions were determined using ANSYS HFSS software by varying the ϵ_r inside the inner layer of the chest model from 1 to 120, its conductivity from 0.3 to 0.5 S/m and loss tangent from 2 to 3. The resultant overdetermined matrix consisting of 480 rows and 36 columns was solved using the LS method to obtain the weight coefficients as demonstrated in Refs. 14,32. To get the ϵ_r -estimates for both simulated and average measured S_{i1} values, the following formula was used based on the simulation model solution:

$$\begin{aligned} \epsilon_r = & 26.66 S_{2,1} - 8.96 S_{3,1} + 3.55 S_{4,1} + 13.87 S_{5,1} \\ & - 7.52 S_{6,1} - 4.88 S_{7,1} + 0.79 S_{8,1} - 4.68 S_{9,1} + 4.74 S_{10,1} \\ & - 9.83 S_{11,1} + 15.11 S_{12,1} - 13.11 S_{13,1} + 11.8 S_{14,1} \\ & + 3.01 S_{15,1} + 12 S_{16,1} - 17.62 S_{17,1} + 7.45 S_{18,1} \\ & - 25.04 S_{19,1} + 20.31 S_{20,1} + 7.53 S_{21,1} - 26.62 S_{22,1} \\ & + 13.77 S_{23,1} + 5.92 S_{24,1} + 0.15 S_{25,1} - 2.09 S_{26,1} \\ & - 0.02 S_{27,1} - 3.64 S_{28,1} - 10.36 S_{29,1} + 14.64 S_{30,1} \\ & - 3.78 S_{31,1} + 0.01 S_{32,1} - 1.32 S_{33,1} + 3.76 S_{34,1} \\ & - 5.96 S_{35,1} + 3.32 S_{36,1} - 6.29 S_{37,1} \end{aligned}$$

(3)

5 | ELECTRICAL PROPERTIES MEASUREMENT USING A DIELECTRIC ASSESSMENT KIT

A Dielectric Assessment Kit (DAK) setup is used to measure the electrical properties (permittivity, loss tangent, and conductivity) of any liquid solution within the created phantom. Figure 5A shows the DAK measurement setup. Glycerin is important for breast imaging. Its permittivity can be varied considerably because it is soluble in water. It is nontoxic and ideal from a human safety perspective.³³ The human chest model with fixed outer and different inner layer electrical properties are formed using a mixture of sodium chloride (NaCl), water, isopropyl alcohol (IPA), and glycerin. The permittivity of the available solutions is measured using the DAK. Figure 5B shows the measured dielectric constant for glycerin, water, and IPA in the frequency range between 10 and 100 MHz. Due to its low dielectric constant, IPA can be used in the inner layer during the experiments for emphysema cases. Different solutions were optimized for normal, edema and emphysema infected human lungs. A 10 mL IPA with 80 mL glycerin solution is used for a normal human lung. For the edema infected lung, a mixture of 200 mL glycerin, 40 mL water, and 10 g NaCl is used. Porcine lung properties are formed using a 200 mL water with 0.7 g NaCl solution. The NaCl–water solution was used to increase the permittivity, electric conductivity, and loss tangent values. The human chest outer layer properties are optimized using 70 mL IPA, 50 mL glycerin, 20 mL water, and 5 g NaCl solution. The obtained dielectric constant values in the

FIGURE 5 Measurement setup of DAK, namely: DAK (A) frequency-dependent characteristics of electric permittivity for: (B) glycerin, water and isopropyl alcohol (emphysema), (C) normal lung, edema, porcine lung, and outer chest layer

FIGURE 6 Measurement setup in the case of normal lung tissue (A). (A) Measurement setup, comparison of average measured and simulated S_{11} for: (B) normal lung, (C) emphysema, and (D) edema models

TABLE 4 Comparison of exact, simulated, and average measured ϵ_r -values for normal human lung and different diseased lung tissue

Category	Exact ϵ_r	Simulated ϵ_r	Measured ϵ_r	Simulated error (%)	Measured error (%)
Normal lung	34.2	33.74	32.99	1.35	3.54
Emphysema	19.5	19.47	18.72	0.17	4
Edema	51.2	51.74	53.67	1.05	4.83

frequency range of 10–100 MHz for all mentioned cases are shown in Figure 5C.

6 | EXPERIMENTAL RESULTS

The values of electric permittivity (ϵ_r) for the normal lung, emphysema, and edema conditions are 34.2, 16.9, and 51.2, respectively, as stated in Ref. 13. The outer layer electrical properties are formed using a mixture of isopropyl alcohol, NaCl, water, and glycerin. A mixture of 1190 mL IPA, 850 mL glycerin, 340 mL water, and 85 g NaCl filled the outer layer of the upper chest torso. The achieved ϵ_r was 40.6 the same as stated in Ref. 15. An 8000 mL glycerin and 1000 mL isopropyl alcohol mixture filled the inner layer of the torso with $\epsilon_r = 34.2$ which is similar to a normal lung tissue. Figure 6A shows the measurement setup for normal lung testing. Port A of the vector network analyzer (VNA) is

connected to the active port and port B is switched to other ports (2–37 ports) for S_{11} measurements. Each port measurement was taken three times and the comparison of the average measured and simulated S_{11} -curves in the case of normal lung tissue are shown in Figure 6B. It is worth noting that averaged measured S_{11} are in good agreement with the simulated ones.

IPA filled the inner layer of the torso with $\epsilon_r = 19.5$ for the emphysema sample. Figure 6C shows the comparison of average measured and simulated S_{11} -distributions for the emphysema test model. A mixture of 7500 mL glycerin, 1500 mL water, and 375 g NaCl solution filled the inner layer of the chest upper torso with $\epsilon_r = 51.2$ which is similar to the edema case. Figure 6D shows the comparison of average measured and simulated S_{11} -curves for edema testing. It is important to note that the averaged measured S_{11} values are matching their simulated ones. Table 4 shows the exact, simulated, and measured ϵ_r -values using Equation (3) with

FIGURE 7 Comparison of average measured and simulated S_{11} for: (A) water and salt mixture (porcine lung), (B) adding 6 water balls, (C) adding 12 water balls, and (D) adding 18 water balls

TABLE 5 Comparison of exact, simulated, and average measured ϵ_r -values for porcine lung with different pulmonary edema stages

Category	Exact ϵ_r	Simulated ϵ_r	Measured ϵ_r	Simulated error (%)	Measured error (%)
Water + salt (NaCl)	108	109.31	110.89	1.21	2.68
Water + salt (NaCl) with 6 water balls	104	104.83	104.9	0.79	0.87
Water + salt (NaCl) with 12 water balls	100	100.3	102.18	0.3	2.18
Water + salt (NaCl) with 18 water balls	96	96.52	98.69	0.55	2.8

an average measured estimation error less than 4.83%. The simulated error is due to LS method weight estimation error.

Another test was conducted for a porcine lung to show the applicability of the sensor under different conditions. The dielectric constant for the porcine lung is equal to 106.8.¹³ The normal water and NaCl solution filled the inner layer of the chest upper torso in ratio: 0.7 g of NaCl in 200 mL of water. The 9000 mL water with 3.15 g of NaCl is used for this case. The obtained ϵ_r -value was 108 which is similar to the porcine lung tissue. Figure 7A shows the comparison of average measured and simulated S_{11} . The average measurement of dielectric constant is close to performed simulation.

To consider different stages of pulmonary edema and making the chest inner layer more nonhomogeneous, small balloons of 200 mL size were filled with water and inserted to this layer. The effective dielectric constant (ϵ_r) of water and NaCl mixture including the water balloons is calculated using³⁴ Equation (4),

$$\sqrt{\epsilon_r} = d \sqrt{\epsilon_{\text{water} + \text{salt}}} + w \sqrt{\epsilon_{\text{water}}} \quad (4)$$

where, d is the volume fraction of water and NaCl solution, w is the volume of water, $\epsilon_{\text{water} + \text{salt}}$ is the dielectric constant of water and NaCl mixture with a value of 108, and ϵ_{water} is the dielectric constant of water with a value of 81. Adding 6 water balls to the inner phantom layer, the obtained average permittivity using Equation (4) was equal to 104. Figure 7B compares the average measured and simulated S_{11} -distributions. The average measurement of the dielectric constant is in good agreement with the simulated one.

When adding 12 water balls in the inner layer, the calculated effective permittivity using Equation (4) was 100. Figure 7C shows the comparison of average measured and simulated S_{11} -distributions. Similarly, adding 18 water balls the comparison of the measured and simulated S_{11} -curves is shown in Figure 7D. The average measurements are matching well with the simulated ones in all analyzed cases. Table 5 shows the exact, simulated and measured ϵ_r -values using Equation (3), for all the cases of a porcine lung tissue with an average measured error of less than 2.8%.

7 | CONCLUSIONS

A 37-port inkjet printed RF based low complexity, low cost, and noninvasive human chest sensor for average permittivity estimation of different lung diseases is presented that can be integrated with cloths easily. The sensor was built on a flexible polyimide substrate of thickness 50 μm . It was found that the

sensor can estimate the dielectric constant for a human lung (normal, edema infected, and emphysema infected) with maximum average error of 4.83%. To differentiate between different stages of pulmonary edema, the inner layer of an elliptical upper torso model was filled with water and salt (NaCl) mixture (porcine lung case) and water balls were added to the inner layer with the increment of six water balls. The average error for pulmonary edema stages detection was found to be less than 2.8%.

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