A Wearable Wireless Monitoring System for the Detection of Pulmonary Edema

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Abstract— In this paper we investigate the feasibility of a simple wearable system that can be used at home to detect or monitor excess fluid buildup in the lungs. This is a medical condition referred to as pulmonary edema. A methodology has been developed to computationally emulate human lungs with various levels of fluid in the alveoli. The proposed wearable system is composed of several small wearable antennas located on the chest and back area. The antennas will operate at MedRadio frequency band and will be optimized for signal penetration through the body. The frequency and time responses of the communication channel between these antennas for the lung models with varying levels of fluid have been measured and analyzed. The results show a correlation between the channel response and the level of fluids inside the lungs. This correlation can potentially be exploited by a simple wearable system to predict the onset of pulmonary edema for patients living in remote areas or people who need to be continuously monitored.

Keywords- Pulmonary Edema, Computational human body models, Wearable antennas, Scattering parameters

I. INTRODUCTION

Pulmonary edema is a medical condition caused by accumulation of excess fluid in the human lungs [1]. There are numerous tiny air sacs (known as alveoli) at the very end of all air tubes branches in the lungs (bronchioles) where oxygen is exchanged with carbon dioxide during the breathing process. The addition of fluid in these air sacs will make breathing difficult. This fluid can accumulate for many reasons including heart problems, pneumonia, exposure to certain toxins and medication, as well as viruses such as COVID-19. Pulmonary edema is a common condition in elderly with about 1 in 15 people aged 75-84 and just over 1 in 7 people aged 85 years and above [1]. If this condition is not treated in time, shortness of breath sets in, leading to acute respiratory distress syndrome (ARDS) which is a form of lung failure. X-ray, CT scan, and Electrical Impedance Tomography are typical technologies that are used to detect and measure fluid build-up in the lungs; however, physical presence and access of the patient to medical/healthcare facilities having necessary equipment are required for accurate diagnosis [1]. Frequent access to such facilities might not be easily feasible or convenient for patients living in remote areas or people who need to be continuously monitored. In addition, during pandemics, resources at the medical facilities are typically diverted to other critical tasks. Therefore, it is desirable to cut down the number of unnecessary visits by the patients during those times.

The recent COVID-19 pandemic has prompted researchers to further explore how the Internet-of-Things (IoT) technology can be efficiently applied to health monitoring and telemedicine. In addition, proliferation and consumer adoption of wearable devices have created a fertile environment for new applications that allow for some health monitoring functions to take place at the patient's home. To the best of our knowledge, there are no wearable or mobile devices currently in the market that can be used to detect fluid buildup in the human lungs.

It is observed that the dielectric properties (i.e., conductivity and permittivity) of fluids like water or saline accumulating in the alveoli are very different than the normal lung tissue or the air inside the bronchial tubes. This is especially noticeable for the frequency range 401 to 406 MHz referred to as Medical Device Radiocommunications Service (MedRadio). MedRadio spectrum can be used for diagnostic and therapeutic purposes in implanted and wearable wireless devices. At these frequencies, the composition of the lungs with accumulated fluid would provide a different effective dielectric environment (or equivalently a communication channel) for a wireless signal passing through the lungs.

In this paper, we investigate the feasibility of a simple wearable technology that can be used at home to detect or monitor fluid buildup in the lungs. A methodology has been developed to computationally emulate human lungs with various levels of fluid in the alveoli. The virtual experiment will use these lung models as part of a full 3D computational human body model that has been used for a variety of research related to applications of wearable, implants and ingestible electronics [2, 3, 4]. The proposed wearable system is composed of several small wearable antennas located on the chest and back area. The antennas will operate at MedRadio frequency band and will be optimized for signal penetration through the body. The frequency and time responses of the communication channel between these antennas for the lung models with varying levels of fluid have been measured and analyzed using the

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computational body area networking platform at the Information Technology Laboratory of NIST. The objective of this study is to show a correlation between the channel response and the level of fluids inside the lungs.

The rest of this paper is organized as follows. Section II describes the 3D computational models that are used to emulate accumulation of fluid in the lungs. Section III describes the antennas that have been designed and optimized for this application. Section IV provides simulation results of the wireless channel (in frequency and time domain) for various densities of the fluid in the lungs. Finally, conclusions and our plans for future work are described in section V.

II. Simulation Platform

To conduct our study, a 3D computational human body model with a resolution of 2 mm has been used. The model includes frequency-dependent dielectric properties of 300+ parts of a male human body. These dielectric properties are user-definable in case custom modifications or changes are desired. Although extensive computational time and complexity often create an obstacle in performing sophisticated simulation involving wearables and implants, the use of computational human body models can effectively capture details of the inhomogeneous environment between the antennas. As shown in Fig. 1, the transmitting and receiving antenna pair are placed on the chest and the back area (just below the shoulder blade). With these locations, there will be an in-body propagation path that goes through the thickest lobe of the right lung. Although, there is always possibility of signal leakage around the torso, a properly designed antenna will radiate most of its RF energy through the body, hence covering primarily the lung tissue.

Although simultaneous use of multiple pairs of antennas to cover both lungs as well as various lobes of each lung could provide more information, in this study we are concentrating on one pair of antennas. Using multiple antennas for the left and right lungs are most likely needed in practice as fluid accumulation might occur only in one side. Here, the transmitting antenna is placed on the right lung since it is bigger in size compared to the left lung [6].

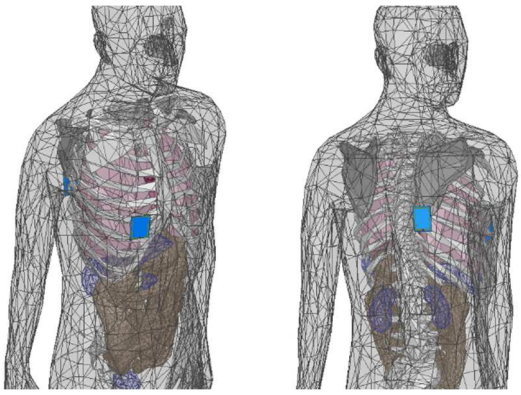

Fig. 1: Computational human body model & the antenna placements

As mentioned in the introduction section, the bronchial tree (i.e. network of alveoli, bronchioles, and bronchi) is responsible for the oxygen exchange with carbon dioxide during the breathing process. In pulmonary edema, the alveoli are partially filled with fluid which impact the patient's breathing. On average, there are 480 million of alveoli in the human lungs, and the mean size of a single alveolus is estimated to be around 4.2 $\times 10^6$ μm^3 [7].

Computational modeling of such small structures within the lung tissue is nearly impossible. However, such microscopic level of modeling might not be necessary for the purpose of our study. The net/accumulating effect of fluid buildup in the air sacs/alveoli is the change in the dielectric properties of the lung's tissue (i.e., the medium where the electromagnetic waves travel through). However, as knowledge of these dielectric properties for various levels of fluid is not currently available, we propose a macroscopic level of computational modeling where a group/cluster of alveoli are considered as one macro-alveolus. As such, the following methodology is used to construct a lung model incorporating the effect of fluid buildup. A periodic lattice of small cubes is created inside the volume of the lungs. Each cube (hereafter referred to as Lattice Element (LE)) represents the overall amount/volume of fluid that has been accumulated in a macro-alveolus. To remove the periodicity of the lattice, an algorithm has been developed to randomize the spacing between the LEs. It should be noted that the actual shape of the lattice elements (i.e., cubes) will not have any impact on the final results and other shapes such as spheres may also be used.

Figure 2 shows a sample of the lung model including the randomized lattice structure. The two parameters in this lung model are the LE size and the average spacing between two adjacent LEs. The average spacing will determine the total number of LEs that are contained within the volume of the lungs. The number of LEs along with their size will also determine the total volume of the LEs that exist inside the lungs. This volume indicates the total fluid buildup which impacts the overall mass density of the lungs representing pulmonary edema. This mass density can be calculated according to the following formula:

$$\rho_{lung}^{(LE\ size,\ LE\ number)} = \frac{\rho_{lung}^{(0,\ 0)} \times (V_{lung} - V_{fluid}^{(LE\ size,\ LE\ number)}) + \rho_{fluid} \times V_{fluid}^{(LE\ size,\ LE\ number)}}{V_{lung}}$$

where

$\rho_{lung}^{(LE\ size,\ LE\ number)}$ = Mass density of the lungs with excess fluid
$\rho_{lung}^{(0,\ 0)}$ = Mass density of the normal lungs tissue (fully inhaled)
ρ_{fluid} = Mass density of the fluid inside the lungs
$V_{fluid}^{(LE\ size,\ LE\ number)}$ = Total volume of the fluid inside the lungs
V_{lung} = Total volume of the lungs.

The substance of the fluid that builds up inside the alveoli is assumed to be 0.9% saline solution. Therefore, $\rho_{fluid} = \rho_{Saline}$ = 1.0046 g/cm^3. Electrical properties of lossy liquids such as sodium chloride (NaCl) are very much temperature dependent. Using the polynomial equations in [8], we can calculate conductivity and permittivity of sodium chloride

solutions as a function of normality and temperature. The temperature range where the model in [8] was validated with actual data was 5-35 °C. However, assuming that the model is still valid for 37 °C (i.e., the normal human body temperature) and using a concentration/molarity of 0.154 mol/L for sodium chloride solutions, the dielectric properties of 0.9% saline solution at body temperature (37°C) can be estimated to be 72.47 F/m in permittivity and 1.86 S/m in conductivity.

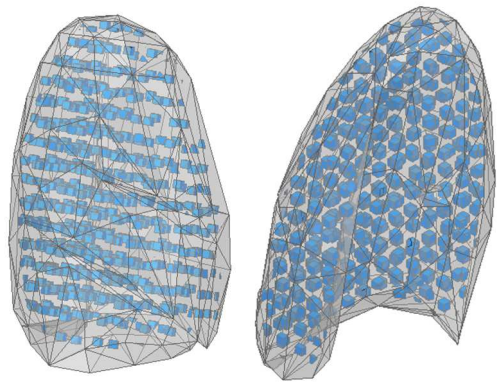

Fig. 2: Computational model of the lungs with a lattice of macro-alveolus

The average mass density of the normal (fully inhaled) lung tissue (ρ_{tissue}) is known to be 0.2 g/cm^3 [9]. Assuming a dimension of 6 mm for an individual cubic-shaped macro-alveolus/LE (equivalent to a volume of 216 × 10^9 μm^3) and considering that the volume of the 3D computational lung model in our simulation is V_{lung} = 2309760.59 mm^3, the following table shows the resulting mass densities of the lungs for various concentration/number of macro-alveolus/LEs. This mass density is used as a parameter to study the impact of fluid accumulation on the forward transmission coefficient of the wireless channel between the two antennas shown in Fig. 1. The possible range of this mass density parameter is from 0.2 g/cm^3 (indicating healthy lung tissue, fully inhaled) to an extreme 1.0 g/cm^3 where the whole lung volume is basically considered to be filled with 0.9% saline solution. Table 1 only shows the relevant information up to 0.45 g/cm^3. It is assumed that the detection of variations in mass densities in the lower range (i.e., up to 0.45 g/cm^3) are more important/critical and should be the main focus for home/remote monitoring applications. Beyond that the patient should most likely transfer to a hospital or clinic to receive proper care.

Table 1: Mass density of lungs for various numbers of macro-alveolus

Average LE Spacing (mm)	Number of LEs	V_{fluid} (mm^3)	$\rho_{lung}^{(LE\ size,\ LE\ number)}$ (g/cm^3)
-	0	0	0.2
9.5	604	105791.19	0.24
5.5	1495	263640.72	0.29
4	2284	400857.96	0.34
3	3173	552898.45	0.39
2.3	3895	678601.29	0.44

III. Antenna

Fig. 3 shows the rectangular loop antenna that has been designed for this application. It operates at the MedRadio band [11]. The loop is sandwiched between two 0.005" (0.13mm) thick Rogers 3010 substrates and the overall size is approximately 5 cm by 5 cm. This size is quite acceptable for a wearable antenna specially since the substrate is thin and flexible.

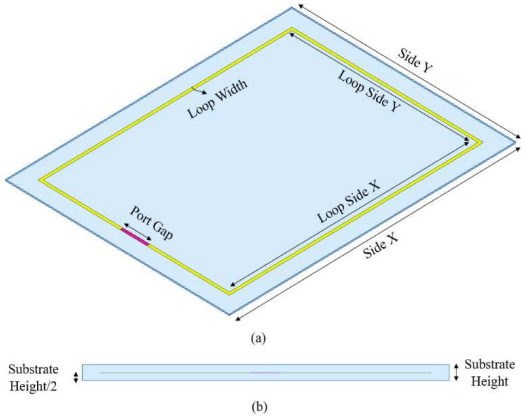

Fig. 3: Rectangular loop antenna (a) isometric view (b) front view

The antenna is placed directly on the human chest & back areas and is expected to conform to the body surface. Since in practice, it could be placed at slightly different positions on the chest/back areas (considering individuals with different body shapes & sizes), the effective permittivity of the environment exposed to the near field of the antenna could also vary [12]. For the same reason, the optimum dimensions of the antennas that are expected to be placed on the chest and back areas are slightly different. This information is shown in Table 2.

Table 2: Dimensions of the receiver and the transmitter antennas

Parameters	Transmitter Antenna (mm)	Receiver Antenna (mm)
Side X	51.4	53.1
Side Y	43.2	41.5
Loop Side X	49.4	47.1
Loop Side Y	37.2	35.5
Loop Width	0.5	0.5
Port Gap	5	5
Substrate Height	0.26	0.26

Since the loading effect is unpredictable to some extent, the antenna substrates add a layer of controlled lossless medium in the near field in order to minimize this unpredictable detuning. The return loss displayed in Fig. 4 demonstrates this further. As observed, the matching does not deteriorate with the changing density of the lungs.

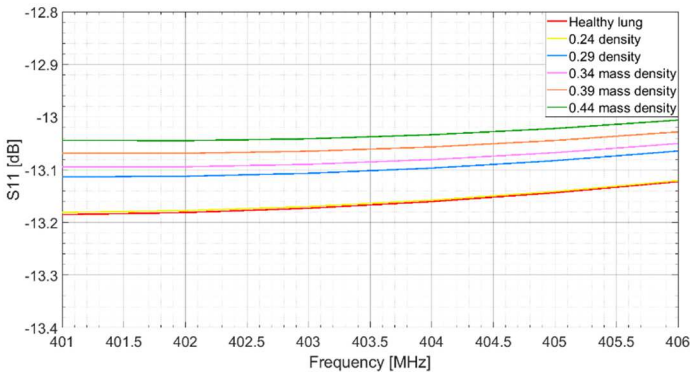

Fig. 4: Reflection coefficient (S_{11}) of the antenna for various mass densities

IV. Simulation Results

Using a pair of antennas described in the previous section and placing them across the right lung as shown in Fig. 1, the forward transmission coefficient (S_{21}) of the channel can be measured for various parameters of the model or the mass densities of the lung shown in Table 1.

We conjecture that the size of the macro-alveolus does not significantly impact the S_{21} results. The potential validity of this conjecture is shown in Fig. 5 where the S_{21} results for a few computationally feasible sizes of the macro-alveolus are presented. As observed, there are minor changes in the S_{21} when the dimension of the modeled macro-alveolus reduces from 6 mm to 5 mm and then 4 mm. It should be noted that while the size of the macro-alveolus is reduced in these simulations, their numbers have been increased in order to keep the total fluid volume (or equivalently the mass density) the same in all three cases. The results shown in Fig. 5 consider a mass density of 0.24. To speed up the execution time of our simulations, a macro-alveolus size of 6 mm size has been considered for the rest of the results presented in this section.

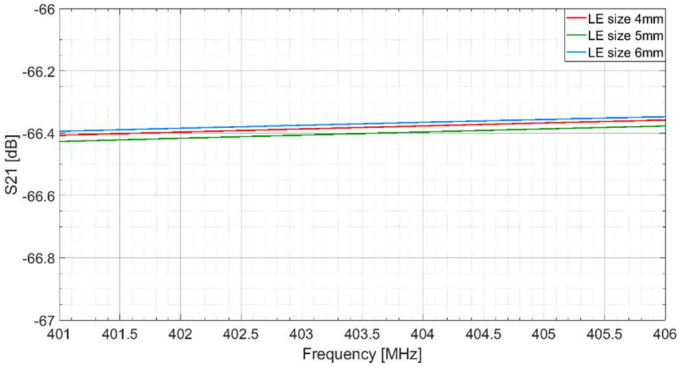

Fig. 5: Forward transmission coefficient (S_{21}) for various LE sizes at fixed mass density

Fig. 6 displays the S_{21} results for various mass densities of the lung ranging from 0.2 g/cm^3 (healthy lung) to 0.44 g/cm^3. As observed, with increasing mass density, the forward transmission coefficient of the channel is gradually reduced. Fluid accumulation in the lung will change the effective dielectric properties of the lungs, which in turn impacts the scattering parameters of the wireless link between the antennas. A noticeable 4.5 dB loss in S_{21} is observed when the mass density increases to 0.44. This change can be easily measured and serve as a warning sign for the onset of the pulmonary edema.

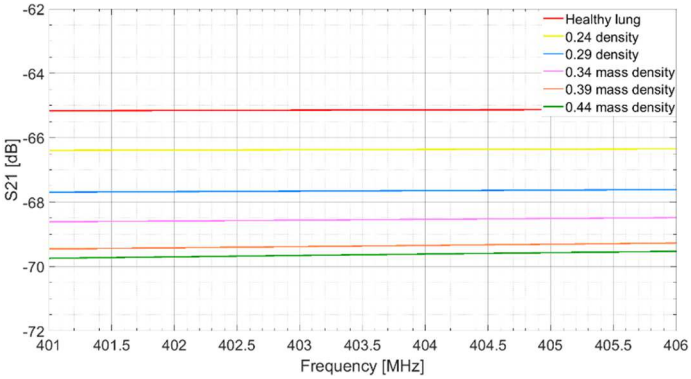

Fig. 6: Forward transmission coefficient (S_{21}) for various mass densities

Time domain responses of the channel have also been obtained using the Inverse Fourier Transform (IFT) after applying a Hann windowing on the S_{21} results. Figure 7 shows these channel responses when an envelope detector is used to remove the mid-band carrier frequency of 403 MHz from the IFT output. Similar to the frequency domain, the impact of the change in the lung mass density is observed through the peak amplitude of the channel responses. The reduction of this amplitude indicates an increase in the lung mass density due to excess fluid accumulation. Therefore, through constant monitoring of the amplitude of the time domain channel response, the onset of fluid buildup in the lungs can be detected. It should be noted that only the relative reduction of the peak amplitude with respect to the baseline response of the channel (i.e., response of the healthy lung) is important. This is because the absolute numerical value of the baseline response could be different for people with different body sizes and shapes. This monitoring of the channel response can be done by the patients at their homes or remotely if additional wireless connection to the Internet is also provided.

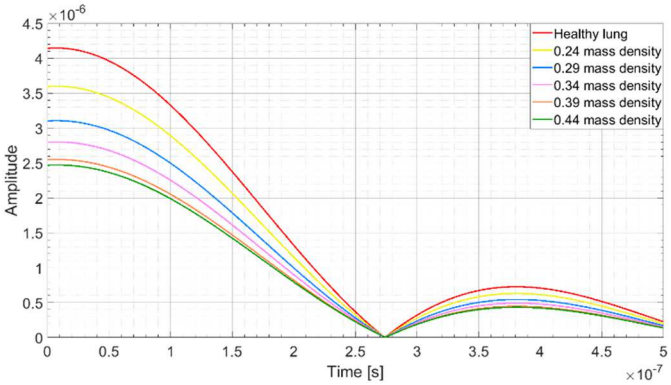

Fig. 7: Time Response of the channel for various mass densities

V. Conclusions and Future Work

The current pandemic climate has significantly increased the importance of IoT technologies that can enable remote or self-health monitoring. The results presented in this paper demonstrates the feasibility of a simple wireless wearable device for detecting pulmonary edema. The main concept used in this research could also serve as a starting point for future research on wearables devices that can assist in detecting abnormalities inside the human organs or tissues.

It is envisioned that the flexible antennas used in this research along with their corresponding electronics could be integrated in the fabric of a wearable band or even a shirt. The shirt/band can be worn at home by patients at risk or individuals at remote areas with no convenient access to a medical facility or hospital. The information provided by the wearable device can also be transmitted through a telemedicine network; therefore, reducing the frequency of unnecessary trips to medical facilities. This would be especially important during a pandemic lockdown.

The study in this paper has been conducted with a static computational model of the lungs. A dynamic simulation platform that can continuously measure the signal during natural inhalation and exhalation could be quite challenging. The dielectric properties used for the model assumes fully inhaled lungs. Therefore, additional circuitry might be needed to ensure the measurements are taken at the right instances during the breathing process. The authors plan to develop more detailed computational models corresponding to fully inhaled and exhaled human lungs in order to study possible advantages of performing measurement at each state. The authors also plan to study application of array antennas to create a more directional path through the lungs; and therefore, further enhancing the resolution of the channel responses.

Ultimate verification of the methodology presented in this paper requires physical experiments involving patients that are susceptible to pulmonary edema.

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References

[1] National Guideline Centre (UK). Chronic Heart Failure in Adults: Diagnosis and Management. London: National Institute for Health and Care Excellence (UK); 2018 Sep. (NICE Guideline, No. 106.) 2, Introduction. https://www.ncbi.nlm.nih.gov/books/NBK536089/

[2] Sayrafian, K. , Krhac, K. , Alasti, M. , Yazdandoost, K. and Simunic, D., "A Study of Capsule Endoscopy Orientation Estimation Using Received Signal Strength", IEEE 28th Annual IEEE International Symposium on Personal, Indoor and Mobile Radio Communications - PIMRC, Bologna, https://doi.org/10.1109/PIMRC.2018.8580786

[3] K. Sayrafian, J. Hagedorn, W. B. Yang, J Terrill, "A Virtual Reality Platform to Study RF Propagation in Body Area Networks", IEEE 3rd International Conference on Cognitive Infocommunications (CogInfoCom), Kosice, Slovakia, Dec. 2-5, 2012

[4] Krhac K., Sayrafian K., Noetscher G., Simunic D. "A Simulation Platform to Study the Human Body Communication Channel," Annual International Conference of the IEEE Engineering in Medicine and Biology Society 2019. 4040-4043. 10.1109/EMBC.2019.8857

[5] A. Jouyban, S. Soltanpour, H-K Chan, "A simple relationship between dielectric constant of mixed solvents with solvent composition and temperature", International Journal of Pharmaceutics 269 (2004) 353–360

[6] Molina D.K., DiMaio V.J., "Normal organ weights in men: part II-the brain, lungs, liver, spleen, and kidneys." Am J Forensic Med Pathol. 2012 Dec;33(4):368-72. doi:10.1097/PAF.0b013e31823d29ad. PMID: 22182984.

[7] M. Ochs , J. R. Nyengaard , A. Jung , L. Knudsen , M. Voigt , T. Wahlers , J. Richter , H. Jørgen G. Gundersen, "The Number of Alveoli in the Human Lung", The American Journal of Respiratory and Critical Care Medicine, Volume 169, Issue 1, Sept. 20, 2003, PubMed: 14512270

[8] Peyman A, Gabriel C, Grant EH. Complex permittivity of sodium chloride solutions at microwave frequencies. Bioelectromagnetics. 2007 May;28(4):264-74. doi: 10.1002/bem.20271. PMID: 17203479.

[9] J. Van Dyk, T.J. Keane, W.D. Rider, "Lung density as measured by computerized tomography: implications for radiotherapy", International Journal of Radiation Oncology, Biology, Physics. 8(8):1363-72, Sept. 1982., doi: 10.1016/0360-3016(82)90587-9.

[10] Garnett E.S. et al. "Lung density: clinical method for quantitation of pulmonary congestion and edema." Can Med Assoc J. 1977 Jan 22;116(2):153-4. PMID: 608146; PMCID: PMC1878992.

[11] Balanis, Constantine A. Antenna Theory: Analysis and Design. 3rd ed. John Wiley, 2005.

[12] S. Dumanli, "Challenges of wearable antenna design," 2016 46th European Microwave Conference (EuMC), 2016, pp. 1350-1352, doi: 10.1109/EuMC.2016.7824602.