NONINVASIVE MICROWAVE Technique
for human cardiorespiratory hemodynamic assessments

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By
Ruthsenne R. Gagarin

Thesis Committee:
Magdy F. Iskander, Chairperson
David Garmire
Hyoung-Sun Youn
Nuri Celik

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ABSTRACT

The use of the noninvasive microwave method for measuring changes in lung water content has been previously reported and validated in animal experiments. The approach is based on measuring the transmission coefficient across the thorax and correlating results with changes in lung water content. Presented in this thesis is the extension of this technique to the monitoring of multiple vital signs including heart rate and breathing as well as the changes in lung water content from a single transmission and reflection coefficient measurement with an electromagnetic coupler. Using a short time Fourier Transform based DSP method, it is shown that these vital signs can be accurately detected and extracted from a single microwave transmission and reflection coefficient measurement. Furthermore, new electromagnetic coupler designs were developed to increase the sensitivity for vital sign detection and additional clinical applications.

A LABView based GUI was developed and utilized in the experimental measurements on a thorax phantom model, and the obtained data confirmed the validity and accuracy of the proposed approach. The radiation bio-safety aspect of this approach was evaluated using a DASY4 near-field scanner, often used for certifying cell phones. Obtained results for a 30mW input power show that the Specific Absorption Rate (SAR) values are about one third of the FCC RF safety standard at the operating frequency of 915 MHz. The electromagnetic coupler design together with the safety experimental procedure and the extracted vital signs results, were submitted to the Institutional Review Board (IRB) and was approved for human clinical trials.
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<tr>
<td>AI</td>
<td>Artificial Intelligence</td>
</tr>
<tr>
<td>BP</td>
<td>Blood Pressure</td>
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<tr>
<td>BrPM</td>
<td>Breaths Per Minute</td>
</tr>
<tr>
<td>CO</td>
<td>Cardiac Output</td>
</tr>
<tr>
<td>DAQ</td>
<td>Data Acquisition</td>
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<tr>
<td>dB</td>
<td>Decibel</td>
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<tr>
<td>DSP</td>
<td>Digital Signal Processing</td>
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<tr>
<td>EKG, ECG</td>
<td>Electrocardiogram</td>
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<tr>
<td>EMD</td>
<td>Empirical Decomposition</td>
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<tr>
<td>FCC</td>
<td>Federal Communications Commission</td>
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<tr>
<td>FEM</td>
<td>Finite Element Method</td>
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<tr>
<td>GHz</td>
<td>Gigahertz</td>
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<tr>
<td>GUI</td>
<td>Graphical User Interface</td>
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<tr>
<td>HCAC</td>
<td>Hawaii Center for Advanced Communications</td>
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<tr>
<td>HF</td>
<td>Heart Failure</td>
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<td>HHT</td>
<td>Hilbert-Huang Transformation</td>
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<td>HR</td>
<td>Heart Rate</td>
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<td>ICG</td>
<td>Impedance Cardiography</td>
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<tr>
<td>ICU</td>
<td>Intensive Care Unit</td>
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<tr>
<td>IEEE</td>
<td>Institute of Electrical and Electronics Engineers, Inc.</td>
</tr>
<tr>
<td>JABSM</td>
<td>John A. Burns School of Medicine</td>
</tr>
<tr>
<td>kg</td>
<td>Kilogram</td>
</tr>
<tr>
<td>kHz</td>
<td>Kilohertz</td>
</tr>
<tr>
<td>LA</td>
<td>Left Arterial</td>
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<tr>
<td>LNA</td>
<td>Low Noise Amplifiers</td>
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<tr>
<td>MHz</td>
<td>Megahertz</td>
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<tr>
<td>mW</td>
<td>Milliwatt</td>
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<tr>
<td>NaCl</td>
<td>Sodium Chloride</td>
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<tr>
<td>NI</td>
<td>National Instrument</td>
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<tr>
<td>PA</td>
<td>Pulmonary Arterial Pressure</td>
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<td>PAC</td>
<td>Pulmonary Artery Catheter</td>
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<tr>
<td>PCP</td>
<td>Primary Care Providers</td>
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<tr>
<td>Pulmonary Edema</td>
<td>Fluid in the Lungs</td>
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<td>RF</td>
<td>Radio Frequency</td>
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<td>S11</td>
<td>Reflection Coefficient</td>
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<td>S21</td>
<td>Transmission Coefficient</td>
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<td>SAM</td>
<td>Specific Anthropomorphic Mannequin</td>
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<tr>
<td>SAR</td>
<td>Specific Absorption Rate</td>
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<tr>
<td>STFT</td>
<td>Short-time Fourier Transform</td>
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<tr>
<td>SV</td>
<td>Stroke Volume, Stroke Volume</td>
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<tr>
<td>TEB</td>
<td>Thoracic Electrical Bioimpedance, Thoracic Electrical Bioimpedance</td>
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<tr>
<td>W</td>
<td>Watt</td>
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<td>WT</td>
<td>Wavelet Transform</td>
</tr>
<tr>
<td>$\varepsilon$</td>
<td>Dielectric Constant</td>
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<tr>
<td>$\sigma$</td>
<td>Conductivity [S/m], Conductivity [S/m]</td>
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Chapter 1 INTRODUCTION

1.1 Background/Motivation

According to the American Heart Association, heart failure (HF), a condition in which the heart can't pump enough blood throughout the body (National Institutes of Health, 2011), affects nearly 5.7 million Americans of all ages and contributes to 300,000 deaths each year. In the U.S. alone, one in every five is at risk for developing HF and 23 million people worldwide. As seen in Figure 1.1, the estimated cost of HF in 2010 was $39.2 billion and 53% of the costs come from hospitalization bills (Lloyd-Jones D, Feb. 2010). Early detection of heart failure is essential for optimal management and prevention of further complications. Coincidentally, higher incidence and prevalence of HF are observed for population with lack of access to healthcare and prior preventive health care. Generally, the mortality following hospitalization for patients with heart failure is 10.4% at 30 days, 22% at 1 year, and 42.3% at 5 years, despite marked improvement in medical and device therapy (Dumitru, Baker, & Ooi, 2011).

Furthermore, the situation worsens due to the expected shortage of physicians by 2020 because of the aging workforce (U.S. Dep’t of Health & Human Serv, 2008) and the disparity between patient to primary care providers in rural areas compared to urban areas are far greater. The PCP to patient ratio in rural areas is 105:100,000 in contrast to the 65:100,000 in urban areas as Figure 1.2 indicates. Residents in rural areas travel an average distance of 60 miles to their physician’s office compared to half the distance for urban consumers (World Health Organization, 2011).
Moreover, when the heart fails to pump enough blood throughout the body, manifestation of many other costly medical conditions such as Pulmonary Edema or fluid in the lungs occurs. In the event of HF, the left ventricle in the heart isn't able to pump out enough of the blood it receives from the lungs. As a result, pressure increases inside the left atrium and then in the veins and capillaries in the lungs, causing fluid to be pushed through the capillary walls into the air sacs as seen in Figure 1.3 (Mayo Clinic, 2011).
Figure 1.3: Pulmonary Edema or fluid in the lungs early symptoms of HF, as a result of fluid build-up when the heart fails to pump the oxygenated blood out that it receives from lungs (Mayo Clinic, 2011).

However, these conditions in addition to other symptoms such as fatigue and shortness of breath are good precursors and indicators of an imminent HF. Therefore, access to health care providers and ambulatory pre-hospitalization devices for early diagnosis and treatments of these conditions will significantly reduce the occurrence of HF. With these staggering statistics, the motivation to mitigate HF along with other heart related diseases, needs no further explanation and calls for urgent, innovative solutions.

One of the solutions that are commonly suggested by health organizations is to transform medical care by how it is delivered or how people access medical services. Advancement in communication and information technology has made this possible through telemedicine or e-medicine. As part of an ongoing research project, our group, Hawaii Center for Advanced Communications (HCAC), is developing a complete wireless tele-healthcare system that includes commercially available sensors such as EKG, breathing, heart rate, and hearing tests for monitoring remote patients and school students. The multi-sensor system is integrated using a LABView based GUI (N. Celik, 2010), and also includes a digital stethoscope and video camera for monitoring the screening procedure and for face to face discussion between doctors and patients. In addition, the system designed for monitoring school students additionally includes hearing and vision tests, once again using commercially available sensors and components. This interactive telemedicine system includes the innovative microwave
sensor which has been successfully used for monitoring changes in lung water content [8-13].

![HCAC's wireless tele-healthcare system](image)

Figure 1.4: HCAC’s wireless tele-healthcare system that includes commercially available sensors such as EKG, breathing, heart rate, and hearing tests for monitoring remote patients and school students. The multi-sensor system is integrated using a LABView based GUI [7], and also includes a digital stethoscope and video camera for monitoring the screening procedure and for face to face discussion between doctors and patients.

It is shown that by measuring the on/across the thorax reflection/transmission coefficients through a unique electromagnetic energy coupler (Iskander & Durney, 1980; M. F. Iskander a. C., Microwave method of measuring changes in lung water, 1983; M. F. Iskander a. C., Electromagnetic techniques for medical diagnosis: A review, 1980; M. F. Iskander C. D., 1982; M. F. Iskander a. C., An electromagnetic energy coupler for medical applications, 1979; Magdy F. Iskander, 1980), sensitive and reliable indication of changes in lung water content in addition to multiple human vital sign parameters such as respiration rate and heart rate can be obtained.

The end goal of the proposed technology will be to one day replace the multiple vital signs sensors and integrated with wireless data transmission technology and has the
potential to enable enhanced chronic care continuous health delivery paradigms, and a more efficient healthcare system in the following domains:

Comfort:

It will facilitate remote continuous monitoring of patients complementing remote triage for high cost and disruptive referral and travel to remote centers.

Healthcare Efficiency:

Reduced number of medical facility visits and hospital stays. By continuous monitoring of several vital signs, some anomalies can be examined and diagnosed without scheduled or unscheduled visits to medical facilities.

Early Detection:

Monitoring changes in vital signs, together with lung water and cardiac indices, continuously allows medical conditions such as congestive heart failure to be monitored, and to assess responses to therapeutic interventions. In addition, early diagnosis of a variety of conditions may be possible in early stages.

Detection of rare, irregular, or evanescent events:

Longer term continuous monitoring could eliminate some hospital observation admissions, augment current home monitoring programs, and prolong the interval of monitoring which is currently available for
ambulatory or home-bound patients, and enable early hospital discharge for patients who require ongoing monitoring.

Enhanced data analysis:

Integration of real-time continuous vital sign trends with other clinical factors is utilized in decision support artificial intelligence (AI) systems in hospitalized patients. Simple ambulatory accurate continuous vital signs monitoring will enable expansion of AI approaches including multi-agent systems and even intelligent agents in chronic disease management paradigms.

1.2 Research Objective

Presently no technology exists that provides continuous, portable, low cost, ambulatory, accurate, and artifact free measurements. This research study is to evaluate and further develop a new technology that would meet all of these goals.

Based on studies [8-13] conducted by Iskander et. al, one device based on microwave method indicated promising results to detect pulmonary edema or fluid in the lungs. Their experimental results on animal lungs and isolated lungs showed correlation between changes in lung water content and left arterial (LA) pressure. The microwave method was made feasible by an electromagnetic applicator designed to couple energy to the body. The energy transmitted is utilized to detect the changes in the dielectric properties of the lungs. The following chapter will provide a brief overview of the microwave method and the experiments conducted in [8-13]. These experiments will provide a basis to the protocols used in this thesis to develop a device to detect additional
parameters for hemodynamic assessments that will essentially be utilized in clinical settings.

Consequently, the focus of this thesis is to improve the sensitivity of the applicator and address challenges for clinical implementation so it may be used for hemodynamic assessments. This thesis proposes two ways to increase the applicator’s efficiency to detect vital signs, one is through improving the design of the applicator and the other is by implementing a novel signal extraction algorithm to separate the acquired microwave signals. Numerical analysis using electromagnetic software will be conducted for various applicator designs. Experimental validation will be conducted using human phantom models as well as manikins that emulate the human respiration cycle. The measured signals will correspond with the changes in respiration and heart rate in addition to pulmonary edema or fluid accumulation in the lungs. Safety concerns will also be addressed by quantifying the amount of energy absorption in the body during the transmission of the microwave signals. Lastly, a proposal based on these experiments was submitted to the Institutional Review Board for human clinical trials. The following section will further discuss the content of this study.

1.3 Outline of Thesis

The following chapters will describe basic components as well as recent development in this measurement microwave procedure. This includes an improved applicator design, phantom material and characterization, new DSP approach for signal separation, respiration and fluid accumulation, RF safety and clinical preparation.
Design optimization of the proposed microwave applicator or coupler will be further discussed in Chapter 3. Topics covered will include the basic structure, design parameters and energy coupling feature. Furthermore, CAD simulations of the applicator on various mediums will also be presented and experimentally validated with fabricated applicators.

Chapter 4 will discuss the proposed signal extraction algorithm that will be used to extract vital signs and fluid accumulation in the lungs from the microwave signal. Extracted vital signs will include respiration rate and heart rate. Experimental validation using data from previous animal experiments will also be included.

Further validation of the proposed algorithm will be experimentally tested in the remaining chapters. In Chapter 5, dielectric property measurements and phantom experiments will be conducted using a LabView based Graphical User Interface (GUI). The dielectric property measurement which includes permittivity and conductivity tests will help determine the proper formula to create phantom human tissues. These tissues will be used in a lung phantom model that emulates the human thorax, mainly respiration and pulmonary edema or fluid in the lungs.

Once the applicator is tested and validated for its ability to detect human vital signs, preparation for human clinical trials soon followed. In Chapter 6, safety concerns based on Federal Communications Commission (FCC) regulations for allowable Specific absorption rate (SAR) levels from microwave devices are presented. In addition, results of the SAR level test conducted at the Kyocera Labs in San Diego, California are also included in this section.
Furthermore, through the collaboration between the medical group at Telehealth Research Institute (TRI) at the John A. Burns School of Medicine (JABSOM) and our engineering group at the Hawaii Center for Advanced Communications (HCAC), we were able to conduct and validate the previous phantom thorax-based models on manikins that are closer representation of the human thorax structure and movement. Experimental procedures as well as additional validation of the microwave applicators efficiency to detect fluid accumulation and respiration in a clinical setting will be presented in Chapter 7. Moreover, clinical and wireless implementation of the microwave system will be tested by replacing the network analyzer with a low cost and lightweight receiver. RF interference from other wireless devices such as Bluetooth will also be studied by quantifying the frequency offset that for the interference threshold. Lastly, summary of the experimental results will be presented in Chapter 8 as well as concluding remarks and future outlook of this study.

**Proposed Approach:**
- Develop a novel microwave applicator
- Measure reflection and transmission coefficient
- Use advanced DSP to separate into multiple vital signs and change in lung water content

**Practical Implementation:**
- RF Safety Levels
- Wireless Implementation (Signal Interference)
- Clinical Trials

Figure 1.5: Outline of the proposed approach and diagram of the microwave system that will be implemented in HCAC’s wireless telemedicine system.
1.4 Hemodynamic Assessment

Given by the definition of HF, a condition in which the heart loses its ability to pump enough blood throughout the body, monitoring and treatment of HF is categorized under a term called Hemodynamic Assessment. Hemodynamic Assessment is a terminology used to describe the measurement of blood flow related topics such as cardiac output which includes heart rate and stroke volume. In common clinical practice, hemodynamic assessment often does not occur until after an acute cardiac episode. If an accurate, noninvasive measurement of cardiac output (CO) monitoring were available, acutely ill and surgical patients undergoing major operations such as a coronary artery bypass graft would benefit. In addition, many patients with chronic and comorbid diseases that ultimately lead to the need for major operations and other costly interventions might benefit from more routine monitoring of CO. Eisenberg et al. (1984) (Eisenberg, 1984) have shown that clinician estimation of CO show poor correlation to measured CO values, thus patients are subject to potential misdiagnosis and mistreatment when CO is part of the therapeutic goal.

The most commonly used method of measuring CO in the ICU is highly invasive, utilizing a flow-directed, thermodilution catheter (also known as the Swan-Ganz catheter), which represents significant risks to the patient (Swan et al., 1970) (Swan HJ, 1973). Components of this device are illustrated in Figure 1.6. In addition, this technique is costly (several thousand dollars per procedure) and requires a skilled physician and a sterile environment for catheter insertion. As a result, it has been used only in very narrow strata (less than 2%) of critically ill and high-risk patients in whom the knowledge
of blood flow and oxygen transport outweigh the risks of the method. In the United States, it is estimated that at least two million arterial catheter monitoring procedures are performed annually.

Figure 1.6: Commonly used method of measuring CO and pulmonary edema in the ICU is highly invasive, utilizing a flow-directed, thermodilution catheter (also known as the Swan-Ganz catheter or Pulmonary Artery Catheter - PAC), which requires a skilled physician and a sterile environment for catheter insertion (Swan HJ, 1973).

A noninvasive way to monitor cardiac hemodynamics would provide exceptional clinical value because data similar to invasive hemodynamic monitoring methods could be obtained with much lower cost and no risk. While noninvasive hemodynamic monitoring can be used in patients who previously required an invasive procedure, a larger impact can be made in patients and care environments where invasive hemodynamic monitoring was neither possible nor worth the risk or cost. An example of
this is that full hemodynamic assessment has been shown to be a powerful predictor of short term heart failure events.

Furthermore, monitoring of cardiac output, the product of stroke volume (SV) and heart rate (HR), be important parameters in clinical decision-making with reference to acute blood loss. This is particularly important in remote triage and battlefield applications where hemorrhagic shock is a major cause of morbidity and mortality. In fact, in wounded military personnel, shock usually occurs because of hypovolemia from acute blood loss, thereby making hemodynamic assessment a keystone of management. Subsequent monitoring to screen for ongoing hemorrhage and to assess the efficacy of resuscitation is important in preventing death and mortality in these patients.

In addition to the above clinical applications, hemodynamic monitoring alone and in conjunction with electrocardiographic (ECG) and blood pressure (BP) measurements may be used to express a variety of contractility indices that have a range of applications outside of a purely clinical context. These applications include fitness/exercise physiology assessments, measurements of autonomic function to aid psychophysiological research and general application as a non-invasive research tool.

1.5 Existing Noninvasive hemodynamic assessment technologies and limitations

Currently there are several available devices that may be used to noninvasively monitor cardiac hemodynamic. These are divided into the categories of External chest mechanical signals; Ultrasound techniques; and External chest electrical signals.
Hemodynamic assessment from external chest mechanical signals typically involves measuring the mechanical function of the heart and these technologies include seismocardiography, thoracocardiography, cardiokymography and kinetocardiography. In varying degrees, issues with these systems arise from the size and cost of the technology, the lack of continuous ambulatory monitoring capability, inability to measure stroke volume and in particular, sensitivity to motion artifact.

Hemodynamic assessment due to ultrasound is via echocardiography and this is the most widely used technology for assessment of mechanical function of the heart and ventricular wall motion. Recently a hand-held, battery powered echocardiograph-Doppler instrument has become available and through this stroke volume measurements become possible (Mondillo et al., 2005). Echocardiography is usually the ‘gold-standard’ for hemodynamic monitoring. However this requires the availability of a skilled operator and cannot be used in continuous, ambulatory assessment situations.

1.6 Electromagnetic Techniques

Hemodynamic assessment from external chest electrical signals generally refers to impedance cardiography (ICG). ICG is also referred to as Thoracic electrical Bioimpedance (TEB), and Electrical Impedance plethysmography. ICG method injects high frequency measurement current via two pairs of electrodes across the thorax (Richard L Summers, 2006). Each pair is placed at the edge of the chest, one pair injects the current and the other pair senses the resulting voltages changes. Several ICG-based commercial devices include BioZ, Niccomo, Osypka and Analogic.
Figure 1.7: The ICG method consists of 4 dual sensors/electrodes with 8 lead wires placed on neck and chest. The outer electrodes inject small current into the skin. The current seeks the path of least path resistance, which is the blood filled aorta which changes volume with each heartbeat and is detected by the inner electrodes my measuring corresponding change in the impedance (CardioDynamics, Co.).

However, despite over 50 years of investigation in this device, one of the major controversies surrounding impedance cardiography concerns how well it compares with conventional methods of determining SV and CO, such as thermodilution (Faes et al., 1999) (Faes TJ, 1999). Since the current injected is minimal, much of the signal is attenuated into the highly conductive chest wall and results in a very low magnitude of the impedance signal (1/1000 of the magnitude of EKG signal). As a result, it is highly susceptible to motion artifacts and consequently can only be used on average size or normal patients at rest or very moderate exercise (Manatec Biomedical, 2010).

Therefore, since the feasibility of the microwave method in (R. Gagarin, Noninvasive microwave technique for hemodynamic assessments, 2010) was proven and has the ideal characteristics needed for continuous, noninvasive, ambulatory hemodynamic monitoring device, this study will focus on further developing the
microwave based technology to increase its sensitivity to detect fluid content in the lungs and additional human vital signs.

Chapter 2 LUNG WATER CONTENT MEASUREMENT

2.1 Microwave Measurement Procedure

In [8-13], the microwave method is based on continuous monitoring of the reflection and/or the transmission coefficient to indicate the changes in the permittivity of the lung tissue. This provides an advantage over other methods such as Electrical Impedance cardiography (ICG) or ultrasounds because microwave signals have greater penetration depth, hence providing more resolution and avoid short-circuiting effect to the tissue highly conductive surrounding the lungs. Prevention of short circuiting effect is crucial because it may interfere with the microwave signals that are being transmitted through the thorax.

According to Iskander et. al., an operating frequency of 915 MHz provided a compromise between resolution and attenuation. Higher frequencies provided more sensitivity but decreased in penetration depth. Measurements were conducted using the phase of the transmission coefficient due to a 50-fold increase in sensitivity [8-13]. The proceeding sections will briefly describe the animal experiments which includes induced pulmonary edema in dogs and isolated lung experiments.

2.2 Prior work, Animal and Isolated Lung Experiments
Evaluation of the microwave transmission method to measure fluid accumulation in the lungs was experimentally validated through several *in vivo* experiments on dogs and isolated lungs [8-13]. For the *in vivo* dog experiments, pulmonary edema was induced in the dog’s lungs. During this procedure, a 915 MHz microwave signal was transmitted across the thorax of the dog. Phase of the microwave transmission coefficient was recorded and indicated a direct correlation between the periodic waveform and the respiration cycle [8-13]. Experimental results of the measured phase and Pulmonary Arterial (PA) pressure is illustrated in Figure 2.1 below.

![Figure 2.1: Correlation between the Pulmonary Arterial (PA) pressure and the measured phase of the transmitted microwave signal during the induced pulmonary edema in in vivo dog experiments.](image)

Accuracy of the microwave method was further validated by eliminating the interference from the extrapulmonary structures using isolated dog lung experiments. The induced lung fluid content was quantified by measuring the weight of the lungs and while
recording the phase of the transmitted signal. Figure 2.2 below illustrates the correlation between the lung weight and phase of the transmitted microwave signal and the increase in phase and magnitude of the reflection and transmission coefficient with the increase of water content which is seen in Figure 2.3.

Figure 2.2: The correlation between the lung weight and phase of the transmitted microwave signal.

Figure 2.3: Increasing phase and magnitude of the reflection and transmission coefficient with the increase of water content.
Feasibility of using the microwave transmission methods on these experiments indicated that unlike other devices that were previously discussed, this coupling microwave applicator have the ability to detect and quantify changes in cases of pulmonary edema. Moreover, because it is not susceptible to motion artifacts or movement of the subject [8-13], it presents great potential as a diagnosis tool for hemodynamic assessments such as respiration rate and heart rate.

2.3 Research Issues with Prior Work on the Microwave Method

Conversely, more research is needed to address the applicators performance and efficiency for human clinical challenges such safety concerns and signal interference. Consequently, the objective of this thesis is to improve the design of the applicator and address safety and clinical challenges (artifact free, accurate, continuous, portable, low cost, and ambulatory), which are outlined below.

Artifact free:

- Applicator is relatively large and requires optimization and miniaturization, needed for optimal coupling between the applicator and the body to increase sensitivity and reduce noise from motion artifacts.

Accuracy:

- Used of transmission coefficient measurements requires two and extensive alignment of applicators
- Recent work discovered that measured signals carried vital signs (breathing and heart beat) in addition to the changes in lung water content.
- Need to develop DSP algorithm to separate signals
Continuous, portable, low cost and ambulatory

- Use of large, **heavy and expensive Network Analyzer**, limited clinical and commercial use
- Wireless implementation and RF interference

Chapter 3 ELECTROMAGNETIC COUPLER DESIGN

3.1 Proposed Design of New Electromagnetic Coupler

The purpose of this microwave coupler/applicator is to provide a passive, noninvasive way of taking measurements of fluid content in a human lung. In addition to measuring the fluid content of the lung, the microwave applicator is also capable of measuring respiratory rate and heart rate. The device is able to measure these points of interest by continuously monitoring the reflection and transmission coefficients of the microwave signal transmitted through the thorax of the person under test. Observing the reflection and transmission coefficients enable the device to detect any changes in the permittivity and conductivity of the intervening tissues which characterizes the amount of fluid accumulation in the lung. In this chapter, a number of the key design parameters for the new electromagnetic coupler such as frequency range, application, size, coupling, and sensitivity to motion artifacts will be briefly presented.

3.1.1 Previous Applicator Designs

Design of the microwave applicator is based on a coplanar transmission line which consists of a center conductor separated from a coplanar ground plane on a low permittivity, thin substrate. The coplanar transmission line is fed through small coaxial
cable and the transmission line is terminated by a resistor equivalent to the characteristic impedance \((Z_o)\) of the coplanar transmission line when it is placed on a biological tissue. Equation \# is used to design the coplanar transmission line where

\[
Z_o = (60 \times \pi) \frac{K'}{K \sqrt{(\varepsilon_{r1} + 1)(\varepsilon_{r2} + 1)}} \tag{Equation 1}
\]

where

\[
\varepsilon_{r1} = \text{relative permittivity of region 1}, \quad \varepsilon_{r2} = \text{relative permittivity of region 2}
\]

\[
k = \frac{w}{w+2g} \quad \text{where} \quad w \text{ and } g \text{ are the widths of the center strip and the gap respectively and}
\]

\[
k' = \sqrt{1 - k^2}
\]

\(K\) and \(K'\) are the complete elliptic integrals of \(k\) and \(k'\), respectively

\[
K = \int_{0}^{\pi} \frac{d\Phi}{\sqrt{1 - k^2 \sin^2 \Phi}} \quad K' = \int_{0}^{\pi} \frac{d\Phi}{\sqrt{1 - k'^2 \sin^2 \Phi}}
\]

Dimensions of the coplanar transmission line are illustrated in the figure below.

Figure 3.1: Dimensions of the microwave applicator which is based on coplanar transmission line.

This geometry allows the spread of the EM fields around the transmission line and only radiates when there is a discontinuity along the line such as contact with the thorax. Power is concentrated at the point discontinuity. Using this design, Iskander and Durney [8;13] developed three applicators. One of the applicators were tested on dialysis
patients, problems they encountered included motion artifacts from patient movements and weak signals due to the higher attenuation in human compared to the isolated lung experiments. The last and third applicator that was developed addressed these challenges with adjustments in the design but has not been verified with human clinical trials. Consequently, based on the previous designs illustrated in the figure below, our team developed new applicators to address these problems which will be discussed in the following section.

![Figure 3.2: Existing microwave applicator by Iskander et. al.](image)

### 3.1.2 Impedance and efficiency of energy coupling issues

Based on the surface transmission line (coplanar waveguide) applicator designed in (M. F. Iskander a. C., An electromagnetic energy coupler for medical applications, 1979), a new applicator was fabricated using copper tape placed on a thin flexible plastic material ($\varepsilon_r = 2.9$). The new design allowed flexibility of the applicator to conform to non-planar surfaces such as a human thorax without losing its form while applying less stress on the copper material. Conformability to the human thorax is critical because the coupling characteristic of the applicator was designed to have matching impedance when
coupled to biological tissues. Using similar phantom experiments that were conducted in (M. F. Iskander a. C., Microwave method of measuring changes in lung water, 1983), sensitivity of the modified applicator was evaluated (R. Gagarin, Noninvasive microwave technique for hemodynamic assessments, 2010).

3.1.3 Design parameters

Using the design equation in (M. F. Iskander a. C., Electromagnetic techniques for medical diagnosis: A review, 1980), additional applicators were developed by our group for various mediums (G. C. Huang, 2011). Widths and the gap design for each applicator were adjusted to have matching impedance when the applicator is coupled to the various materials. The first two applicators were designed to be coupled to the human thorax ($\varepsilon_r = 54$). Upon evaluating the applicator designed in, mismatch may occur at the junction between the coaxial cable to the center conductor. This mismatch increased the sensitivity to motion artifacts and prevented the energy from propagating to the rest of the structure. Consequently, epoxy was added on the junction of the original applicator in []. However, the width of the center conductor and the gap between the ground plane and the center conductor was designed to match human muscle and not epoxy. With this in mind, another applicator (App 2) was designed with a modified junction to match epoxy rather than human muscle. Furthermore, since the thoracic region is not a planar surface, reducing the size of the applicator (App 3) in theory would minimize the occurrence of mismatch since the coupling area is less. Using the original with epoxy design as a
baseline, it was scaled down to 50% and extended the ground plane towards the coaxial cable. However, more experiments are needed to quantify the possible reduction of sensitivity to vital signs for smaller structures.

In addition to improving the matching characteristics of the applicator, increasing the bandwidth of the applicator is desirable for wider range of application such as imaging at higher frequencies. As a result, an applicator (App 4) based on the modified junction model was developed by minimizing the gap and the width. Lastly, two applicators (App 5, App 6) were designed to couple to plastic instead of human muscle for the RF Safety experiments which will later be discussed in Chapter 6. The figure below illustrates the schematic of each applicator. Similar to the applicators designed in (R. Gagarin, Noninvasive microwave technique for hemodynamic assessments, 2010), each applicator was fed with a flexible coaxial cable (RG 178) and the transmission line was terminated with a 50Ω lumped resistor.
App4: Broadband
\[ a = 52.36, \ b = 40.8, \ g = 3.8, \ g' = 0.42, \ w = 0.4, \ w' = 4.16 \]

App5: Plastic1
\[ a = 52, \ b = 36, \ g = 2, \ w = 32 \]

App6: Plastic2
\[ a = 52, \ b = 36, \ g = 1.6, \ w = 35 \]

Figure 3.3: Schematic (in mm) of the applicator designs for various mediums and frequency range.

The new applicator designs were numerically evaluated using electromagnetic CAD models on HFSS. Parameters and results of the simulation are discussed in the following section.

### 3.2 HFSS Simulations

#### 3.2.1 Simulation Parameters

Using HFSS which is based on Finite Element Method (FEM), the applicator models from the previous section were simulated. Models and the simulation parameters are included below. Frequency of operation varied from 500 MHz to 10 GHz depending on the type of applicator. There were two sets of layered structure; one for the muscle applicators and the other is for the plastic applicators, both of which are illustrated in Figure 3.5: 3D view, planar layer model, side view and top view of the HFSS simulation setup for applicators designed for human muscles. Thickness of the muscle is 60mm and respectively. A symmetry plane along the YZ axis was included to minimize the
processing time. As a result, the 50Ω lumped resistor at the end of the center conductor was replaced with 100 Ω. The applicator was excited using a wave port placed at the end (farthest away from the applicator) of the coaxial cable. The simulation boundary varied for each applicator. Thickness of the phantom muscle ($\varepsilon_r @ 915\text{ MHz} = 54$, $\sigma @ 915\text{ MHz} = 1.16$) and the plastic ($\varepsilon_r @ 915\text{ MHz} = 2.9$, $\sigma @ 915\text{ MHz} = 0$) was set to 60 mm and 2 mm respectively. Additional simulation setup and parameters as well as the HFSS applicator models are included in the figure below.

Figure 3.4: HFSS simulation setup for S11

Table 3-1: HFSS simulation parameters

<table>
<thead>
<tr>
<th>Parameters</th>
<th>$\varepsilon_r$</th>
<th>$\sigma$ [Sm$^{-1}$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Material Properties</td>
<td>$\varepsilon_r$</td>
<td>$\sigma$ [Sm$^{-1}$]</td>
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<tr>
<td>Air</td>
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<td></td>
</tr>
<tr>
<td>Epoxy</td>
<td>3.9</td>
<td></td>
</tr>
<tr>
<td>Muscle</td>
<td>54</td>
<td>1.16</td>
</tr>
<tr>
<td>PEC</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Plastic</td>
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</tr>
<tr>
<td>Polyethylene</td>
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<tr>
<td>Vacuum</td>
<td>1.0</td>
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<tr>
<td>Simulation Parameters</td>
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<td>Excitation Port</td>
<td>Wave Port</td>
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<tr>
<td>Lumped Element</td>
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<td></td>
</tr>
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<td>Frequency Range</td>
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<td></td>
</tr>
<tr>
<td>Symmetry Plane</td>
<td>YZ</td>
<td></td>
</tr>
</tbody>
</table>
Figure 3.5: 3D view, planar layer model, side view and top view of the HFSS simulation setup for applicators designed for human muscles. Thickness of the muscle is 60mm.

Figure 3.6: Planar layer model, side view and 3D view of the S21 HFSS simulation setup for applicators designed for plastic surfaces. Thickness of the muscle and plastic is 60mm and 2mm respectively.
Figure 3.7: Various applicator models simulated on HFSS (G. C. Huang, 2011).

3.2.2 Simulation Results

Measured impedance and magnitude of the reflection and transmission coefficients are illustrated in the figures below. For S11, the modified junction applicator performed best with -15 dB from 500 MHz to 5 GHz. Furthermore, with an exception to the small modified original applicator, of all the muscle coupling applicators, the S21 of the modified junction applicator fluctuated between -35 dB to -50 dB from 500 MHz to 5 GHz. Although the broadband applicator has a wider range of operable frequency, the modified junction applicator performs better at 915 MHz for S11 and 20 dB better in the lower frequencies and 10 dB better for frequencies higher than 2.4 GHz for S21.
According to the e-field plots in the figures Figure 3.11 and Figure 3.12 below, all of the applicators except the modified small original applicator indicated strong e-field around the coaxial cable. Difference in the design with the modified small original applicator compared to the other applicators was the extension of the ground plane towards the coaxial cable. Extending the ground plane allowed the energy to propagate more throughout the structure rather than reflected back to the cable.

Figure 3.8: Simulation results of S11 on HFSS

Figure 3.9: Simulation results of S21 on HFSS

Figure 3.10: Original w/o Epoxy in the YZ plane and XY plane

App1: Original with Epoxy
App2: Modified Junction
App3: Small original
3.3 Fabricated Prototypes of the EM Couplers for various mediums

Based on the simulations from the previous section, six sets of applicators were fabricated for human thorax and plastic. The applicators were fabricated using Copper tape (3M) and placed on a thin flexible plastic material for backing. The following figure illustrates the fabricated applicators.
Figure 3.13: Fabricated prototypes of various applicators for human muscle and plastic surfaces.
Chapter 4 SIGNAL PROCESSING

4.1 Introduction and Existing algorithms

In addition to the new applicator designs, our group has developed a digital signal processing (DSP) algorithm to individually extract useful vital sign parameters such as breathing and fluid accumulation from the microwave signal and potentially extract heartbeat when used in clinical applications. Similar to the previously mentioned techniques of biological signal measurements such as ECG and ICG, decomposition and extraction algorithms are crucial components to quantifying the trade off of computational efficiency and accuracy of the acquired signals. Some of extraction algorithms of physiological signals include Wavelet Transform (WT) – used by (Akay, Welkowitz, Semmlow, & Kostis, 1993) for characterizing heartbeat sounds, Empirical Mode Decomposition (EMD) and Hilbert-Huang transformation (HHT) to decompose ventricular fibrillation from ECG measurements (M. C. Wu, 2009). This chapter begins by introducing the basic concept of this algorithm and its advantages using computer simulations. This will be followed by an experimental validation of its accuracy using data from previous animal studies.

4.2 Signal Extraction

The proposed DSP algorithm (Celik, Gagarin, Youn, & Iskander, 2011) is based on short-time Fourier Transform (STFT) method and linear regression. STFT and linear regression is implemented to extract the useful vital sign parameters from the integrated (combined) signal obtained from the microwave measurements. Compared to the wavelet
method, the STFT based signal extraction algorithm has much less computational complexity and is easier to implement on DSP chips.

The developed DSP method first divides the data into nonoverlapping windows. Then, a linear regression operation is applied to each window to remove the linear trend due to lung fluid content changes. An STFT is applied to estimate the spectral contents of each window, and then the spectrum averaging process described in (A. H. Madsen, 2008) is applied to increase the signal to noise ratio. From this averaged spectrum, estimates of the vital sign parameters can be extracted by peak detection (A. H. Madsen, 2008). In a typical phase measurement, two peaks are expected to exist in the signal spectrum corresponding to respiration and cardiac output. In signal extraction, the breathing signal is first estimated and then the second peak corresponding to the weaker cardiac output is sought. In order to eliminate the error due to strong harmonics of the breathing signal being detected as the heart signal, the spectrum peak of the heart signal is searched in the frequency range of 0.8 to 3 Hz (A. H. Madsen, 2008). Performance of the developed signal extraction method is evaluated through computer simulations which is illustrated in Figure 4.1. For 10 Hz data sampling rate, a window size of 256 samples with spectrum averaging of 8 windows are selected since these parameters lead to a very high accuracy without excessive estimation delays.
4.3 Experimental Validation on Animal Experiments

In order to validate the accuracy and efficiency of the developed DSP method, measured data from an animal experiment (Iskander & Durney, 1980; M. F. Iskander a. C., Microwave method of measuring changes in lung water, 1983; M. F. Iskander a. C., Electromagnetic techniques for medical diagnosis: A review, 1980; M. F. Iskander C. D., 1982; M. F. Iskander a. C., An electromagnetic energy coupler for medical applications, 1979; Magdy F. Iskander, 1980) was utilized as a benchmark to illustrate its sensitivity to extract the dog’s vital signs such as heart rate which is shown in Figure 5. From this data, a breathing rate of 8 breaths/min is found with a heart rate of 48 beats/min which are within the nominal ranges for a dog under anesthesia. These results indicate that it is possible to extract the heart signal with much smaller amplitude from the rest of the data.

Figure 4.1: The components of composite phase signal.
Application of the algorithm on additional phantom experiments will be discussed in the proceeding sections.

Figure 4.2: The DSP signal extraction results from previous dog experiments (R. Gagarin, Microwave Stethoscope: A New Method for Measuring Human Vital Signs, 2011).
Chapter 5 THORAC-BASED PHANTOM EXPERIMENTS

To experimentally validate the effectiveness of the signal extraction algorithm, phantom experiments similar to the animal lung experiments were performed. The phantom experiments included developing phantom human tissue and emulating human respiration and fluid accumulation in the lungs. Moreover, in order to ensure that the tissue-equivalent phantom had the same electrical characteristics as the human thoracic muscles, dielectric property measurements were conducted. A LabView-based GUI was also developed to acquire the raw data from the Network Analyzer.

5.1 Dielectric Property Measurements

5.1.1 Dielectric Property Measurement Set-up

The dielectric properties relative permittivity ($\varepsilon_r$) and conductivity ($\sigma$) of phantom human tissue over various frequencies were based on the FCC standards (FCC, June 2001). These standards were derived from the work of C. Gabriel et al (Gabriel, 1996) using the summation of 4 cole-cole equations. With an operational frequency of 915 MHz, the relative permittivity ($\varepsilon_r$) and conductivity ($\sigma$) of the muscle tissue are 54.6 and 1.03 S/m respectively.

$$\varepsilon(\omega) = \varepsilon_\infty + \sum_{m=1}^{4} \frac{\Delta \varepsilon_m}{1 + (j\omega \tau_m)^{(1-\alpha_m)}} + \frac{\sigma j}{j\omega \varepsilon_0}$$  \hspace{1cm} (Equation 2)

Where

- $\varepsilon_\infty$ is the permittivity in the terahertz frequency range,
- $\varepsilon_0$ is the permittivity of free space, ($8.854 \times 10^{-12}$) \left[ \frac{F}{m} \right]$. 
\( \omega \) is the angular frequency, \( \omega = 2\pi f \text{(Hz)} \)

\( \sigma_j \) is the ionic conductivity, \( j = \sqrt{-1} \)

\( \tau \) is the relaxation time and

\( \Delta \varepsilon \) is the drop in permittivity

in the frequency range corresponding to \( 1 \gg \omega \tau \gg 1 \)

Using NaCl, a gelling agent called TX151, and de-ionized water \( (\varepsilon_r = 78) \), various muscle mixtures were tested to determine the proper solution that would produce a phantom human muscle tissue. A dielectric probe kit was used to determine the electrical properties of the phantom muscle tissues. Experimental setup of the dielectric measurements are illustrated in Figure 5.1: Agilent Probe Kit or Dielectric Measurement Set-up. It includes a network analyzer (PNA E8364B) and a dielectric probe (Agilent). The dielectric probe was calibrated using de-ionized water, metal block, and teflon. The dielectric properties of these materials are included in Table 5-1

Figure 5.1: Agilent Probe Kit or Dielectric Measurement Set-up
5.1.2 Permittivity and Conductivity Measurements

The percent of NaCl in the solution were adjusted to change the dielectric properties, while the TX151, was adjusted to develop the desired consistency of the phantom muscle. The table below contains the percent and actual weight of the H$_2$O, NaCl and TX150 mixture of a 15 (g) solution. Error! Reference source not found. contains the measured permittivity and conductivity of the phantom tissue samples along with the calibration materials.

Table 5-1: Phantom Muscle Mixture Solution

<table>
<thead>
<tr>
<th>Material</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>H$_2$O</td>
<td>(%)</td>
</tr>
<tr>
<td>NaCl</td>
<td>(%)</td>
</tr>
<tr>
<td>TX150</td>
<td>(%)</td>
</tr>
<tr>
<td>Muscle</td>
<td></td>
</tr>
<tr>
<td>75</td>
<td>10.5</td>
</tr>
</tbody>
</table>

Figure 5.2: Measured dielectric constant ($\varepsilon \approx 53.5$) – left, and conductivity ($\sigma \approx 1$) – right, of the
phantom muscle at 915 MHz compared with the Cole-Cole model ($\varepsilon \approx 52$), ($\sigma \approx 0.95$) respectively.

To further validate the dielectric property of the phantom tissues and performance of the fabricated applicators, a thoracic-based human phantom model was created. Experimental procedures and results of this experiment will be discussed in the following section.

5.2 LABView-based GUI

To conduct the phantom experiments, a LabView based graphical user interface (GUI) was developed. Basis of the GUI is similar to the display and data acquisition capabilities of a network analyzer. Unlike MatLab, LabView is a graphical programming environment used to develop measurement, test and control systems using graphical icons connected by wires, similar to a flowchart. Via a GPIB (NI GPIB-USB-HS) cable, a computer was able to send commands to control the network analyzer and receive specific data using the LabView GUI.

Figure 5.3: Block diagram of LabView, a graphical programming environment that uses graphical icons.

Figure 5.4: The LabView GUI uses a GPIB cable to control and acquire data from the network analyzer.

As illustrated in Figure 5.5 below, parameter (1) may also be changed depending on the mode or hardware that is utilized to acquire or control an instrument. Within the channel
parameters, users can also chose measurements parameters such as reflection and/or transmission coefficients (2). Sweep type such as Linear Frequency or Continuous Wave Time (3) in addition to the span or frequency range of measurement (4). Timing options and number of data points are also incorporated (5, 6). Lastly, a graphical representation of the acquired data (processed or raw) is illustrated in a graphical waveform.

![Graphical Representation](image)

Figure 5.5: LabView based GUI to control the network analyzer for data acquisition and measurements of the reflection and transmission coefficients within the set operating frequencies. Raw or processed data is represented in either tabular or graphical form.

### 5.3 Lung Phantom Experiments

Using the GUI discussed in the previous section, a thoracic-based human phantom utilizing the phantom tissue and applicators from the previous sections will be presented in this section. Following that, the protocol for the human respiration as well as the induced pulmonary edema will be shown. Furthermore, measured parameters such as the magnitude and phase of the reflection and transmission coefficients will be presented as well.

To further validate the dielectric property of the phantom tissues, the magnitude of the reflection coefficient and impedance of the fabricated human applicators were
measured while coupled to the phantom tissue using a network analyzer (PNA E8364B). Set-up of this measurement is included in Figure 5.5 and Figure 5.7 and Figure 5.9 contains the measured magnitude and impedance of S11.

5.3.1 Respiration and Fluid Accumulation Experiment

For the emulation of the respiration and fluid accumulation, a thoracic-based human phantom model was used. To emulate human respiration or breathing, a balloon attached to a manual air pump was inserted between the applicator and sponge. The sponges represent the air sacs in the lungs that would accumulate fluid, similar to a human experiencing pulmonary edema. The fluid was inserted into the sponges using a syringe. The phantom muscle ($\varepsilon_r = 54.9$, $\sigma = .95 \text{ [S/m]}$) (Gabriel, 1996) discussed in the previous section was then placed on outer sides of the sponges. The human applicators were placed between the phantom muscle and the Styrofoam that kept the structure in tacked. Rubber bands around the Styrofoam provided the tension needed to bring the structure together while the air was released out of the balloon. Figure 5.7 and Table 7-2 the set-up and layered planar thorax model respectively. The applicators were connected to a network analyzer (Agilent PNA E8364B) which was used to measure the impedance, phase and magnitude of the reflection and transmission coefficients.
Figure 5.6: Setup of the impedance and magnitude measurement to validate the coupling characteristics of the applicator to the phantom muscle.

Figure 5.7: A photo of the experimental setup of the thorax-based phantom model. The overall dimension of the model is 10cm x 10cm x 10cm.

Table 5-2: Layered planar structure of the phantom thorax model illustrating the microwave applicator placed in direct contact with the muscle layers and secured in position using two blocks of Styrofoam.

<table>
<thead>
<tr>
<th>Material</th>
<th>Styrofoam</th>
<th>Applicator</th>
<th>Phantom Muscle</th>
<th>Sponge</th>
<th>Balloon</th>
<th>Sponge</th>
<th>Phantom Muscle</th>
<th>Applicator</th>
<th>Styrofoam</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness (mm)</td>
<td>20</td>
<td>.066</td>
<td>2-4</td>
<td>20</td>
<td>-</td>
<td>20</td>
<td>2-4</td>
<td>.066</td>
<td>20</td>
</tr>
</tbody>
</table>

As seen in Table 5-3, there were three segments in this experiment, first was to establish a baseline, followed by normalized breathing and lastly, fluid accumulation. The signal baseline was acquired for about a minute. Breathing was initiated with about 12-15 breaths per minute (BrPM). Once the breathing rate was stabilized, fluid was simultaneously injected into the lungs at rate of 2cc per minute. This rate may be adjusted according to the size of the sponges. The following section presents the measured data from these experiments.
Table 5-4: Snap shot of a time stamp of the protocol for phantom experiment

<table>
<thead>
<tr>
<th>Time (sec)</th>
<th>Tasks</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 – 60:</td>
<td>Develop baseline, no breathing, no fluid</td>
</tr>
<tr>
<td>60 – 150:</td>
<td>Breathing cycle of 12-15 BPM</td>
</tr>
<tr>
<td>150 – 210:</td>
<td>Insert 2cc of fluid into lungs in addition to ongoing breathing</td>
</tr>
<tr>
<td>210 – 270:</td>
<td>Insert 2cc, total of 4 cc inserted</td>
</tr>
<tr>
<td>270 – 330:</td>
<td>Insert 2cc, total of 6 cc inserted</td>
</tr>
<tr>
<td>330 – 390:</td>
<td>Insert 2cc, total of 8 cc inserted</td>
</tr>
<tr>
<td>390 – 420:</td>
<td>Insert 2cc, total of 10 cc inserted, Breathing ends</td>
</tr>
<tr>
<td>420 – End :</td>
<td>Fluid insertion ends, Acquire end baseline</td>
</tr>
</tbody>
</table>

5.3.2 Magnitude of the Reflection Coefficient (S11)

Both simulation and experimental data are operational till 5 GHz. At 915 MHz and 2.4 GHz, their magnitude is nearly the same at about -27 dB and -20 dB respectively. One significant difference is the presence of resonance for the experimental data. This is due to the losses and movement from the cable, compared to the ideal case in the simulation result. However, trend line of the experimental result is similar to that of the simulation data.

![Figure 5.8: Comparison of the simulated and measured magnitude of the reflection coefficient.](image)

Resonances are present in the experimental result due to the cable.
5.4 Results and Verification of Proposed DSP Algorithm

Measured phase of S11 and S21 are illustrated in the figure below. Initially, 50 seconds of inactivity was created to develop a baseline for the microwave signal. The baseline indicated a phase change of 4-5 degrees for S21 and 1-2 degrees for S11. After 50 seconds, manual breathing was initiated to emulate a breathing cycle of 4 seconds inhale, 2 seconds hold and 4 seconds exhale. This breathing cycle is equivalent to the ideal breathing cycle of 12-15 BrPM for adults. Phase changes due to breathing were measured at an average of 25-30 degrees for S21 and 5 degrees for S11. At the end of 10 breathing cycles, water was simultaneously injected into the sponges at rate of 2cc every minute to emulate fluid accumulation in the lungs for a total of 10cc. This resulted in 40 degrees and 15 degrees phase changed for S21 and S11 respectively.

![Graphs showing measured phase changes](image)

Figure 5.9: Measured phase changed of the transmission and reflection coefficients due to breathing and fluid accumulation.
After verifying the operation of the STFT based DSP method with the data from the dog experiments, the results of the phantom experiments were processed to obtain the breathing rates and lung fluid contents. The developed DSP method was applied to the data plotted in Figure 5.9: Measured phase changed of the transmission and reflection coefficients due to breathing and fluid accumulation and extracted the respiration and fluid accumulation rate from the phantom experiment from the transmission and reflection data. In Figure 5.11 the measured change in the phase of the transmission coefficient is approximately 50° compared to a 18° change in the reflection coefficient shown in Figure 5.11. Although the changes are significantly smaller in the reflection coefficient measurements, the proposed DSP method successfully extracted the same respiration rate of 0.08 Hz and the changes in the lung water. However, the respiration amplitude turned out to be about 7-10 degrees for the transmission coefficient compared to that of 0.5 degree for the reflection coefficient measurements.

Figure 5.10: Extracted signals from the transmission coefficient from the phantom experiment.
Figure 5.11: Extracted signals from the reflection coefficient from the phantom experiment.
Chapter 6 RF SAFETY

6.1 FCC Regulations

Safety standards, recommendations and guidelines for exposure to radio frequency and microwave energy have been developed independently by a number of international and national organizations including the American National Standards Institute (ANSI) and the IEEE (ANSI/IEEE C95.1-1992). These guidelines have been developed by panels of scientists and medical experts to protect human beings from known harmful levels of exposure to RFEM fields. Based on present knowledge, the IEEE supports the conclusion that exposure at or below the levels recommended in ANSI/IEEE C95.1-1992 is not harmful to human health. In 1997, the Federal Communications Commission (FCC) adopted the guidelines set by ANSI and IEEE and developed a document called OET Bulletin 65 (FCC, June 2001) to provide assistance in determining whether proposed or existing transmitting facilities, operations or devices comply with limits for human exposure to radiofrequency (RF) fields. RF safety measurements that were based on the FCC guidelines of the applicators will be presented in this chapter.

These guidelines assert the following safety considerations:

(a) Exposure to electromagnetic fields at frequencies above about 100 kHz can lead to significant absorption of energy and temperature increases. At frequencies from 10 MHz to 300 GHz, heating is the major effect of absorption of electromagnetic energy, and temperature rises of more than 1–2 °C can have adverse health effects such as heat exhaustion and heat stroke.
(b) The sensitivity of various types of tissue to thermal damage varies widely, but the threshold for irreversible effects in even the most sensitive tissues is greater than 4 W/kg under normal environmental conditions. These data form the basis for an occupational exposure restriction of 0.4 W kg, which provides a large margin of safety for other limiting conditions such as high ambient temperature, humidity, or level of physical activity. This factor of 10 was used to provide a large margin of safety for other limiting conditions such as high ambient temperature, humidity, or level of physical activity.

(c) To measure this, all specific absorption ratio (SAR) values are to be averaged over any 6-min period. Peaks of the SAR can range more than an order of magnitude above a whole body average and thus spatial peak SAR values below 8 W/kg as averaged over any one gram of tissue should also not be exceeded.

Summary of the FCC guidelines for acceptable SAR levels is included in the table below. The definition of SAR will be further explained in the next section.

<table>
<thead>
<tr>
<th>SAR [ W/kg]</th>
<th>FCC Guideline</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Controlled/Occupational</td>
</tr>
<tr>
<td>Whole Body</td>
<td>0.4</td>
</tr>
<tr>
<td>1g. Av (Partial Body)</td>
<td>8</td>
</tr>
<tr>
<td>10g. Av (Hands, wrists, feet and ankles)</td>
<td>20</td>
</tr>
</tbody>
</table>
Specific Absorption Rate (SAR)

Specific-absorption-rate (SAR) measurements are critical for understanding the effects of non-ionizing radiation, such as the emissions from devices that is based on radio frequency (RF) technology on biological tissues. Energy emitted from these devices, such as cell phones, are commonly referred to as electromagnetic or RF radiation. Exposure to very high RF intensities may result in tissue damage due to thermal effects or heating of the tissue. Safety standards for human exposure to RF energy are evaluated by the Specific Absorption Rate (SAR), which quantifies the rate at which energy is absorbed per unit mass in an exposed object. Specific absorption rate or SAR is the time derivative of the incremental energy (dW) absorbed by or dissipated in an incremental mass (dm) contained in a volume (dV) of a given density (ρ):

$$SAR = \frac{d}{dt} \left( \frac{dW}{dm} \right) = \frac{d}{dt} \left( \frac{dW}{\rho dV} \right)$$  \hspace{1cm} (Equation 3)

SAR should be considered an “absorbed dose rate” and is related to electric fields at a point by:

$$SAR = \frac{\sigma |E|^2}{\rho}$$ \hspace{1cm} (Equation 4)

Where:

$\sigma$ = conductivity of the tissue (S/m)

$\rho$ = mass density of the tissue (kg/m$^3$)

$|E|$ = magnitude of the measured electric field (V/m)
To account for near-field energy coupling effects, transmitters are evaluated with electric field measurements inside homogeneous tissue models or computer modeling techniques using anatomically equivalent tissue models.

6.3 SAR Measurements

Validation of the safety compliance were conducted at Kyocera Wireless Labs in San Diego, CA using the DASY4 system (Schmid & Partner Engineering AG – SPEAG, Zurich, Switzerland), an automated near-field scanning system that measures SAR levels on biological tissues. The following subsections will further describe the DASY4 system, SAR measurement procedure and results. Detailed protocol of the SAR measurements is included in the Appendix section under RF Safety.

6.3.1 SAR Equipment – DASY4

Although there are numerous systems used to validate SAR levels, the DASY systems have been the most widely used system in the industry. The DASY4 system consists of the following:

- Measurement server for signal filtering and controls the high precision 6-axis robot
- Computer with Windows XP and DASY4 software
- Dosimetric probe with an optical surface detector system calibrated for use in liquid with high permittivity
- Probe alignment unit
- Data Acquisition Electronic (DAE)
- Electro-optical converter (ECO)
- Specific Anthropomorphic Mannequin (SAM) twin phantom
- Tissue simulating liquid
- Validation dipole kits to verify proper functioning of the system
Components of the DASY4 system are illustrated in Figure 4. The phantom that was used for the SAR measurements is called Specific Anthropomorphic Mannequin (SAM 12) which holds about 25 liters of tissue equivalent liquid. A cover is used to prevent liquid evaporation when it is not being used. The shell corresponds to the specifications in IEEE 1528-2003, CENELEC 50361 and IEC 62209. The shell thickness is $2 \pm 0.2$ mm, and is 1000 mm long and 500 mm wide. Dielectric constant ($\varepsilon_r$) of the phantom shell is less than 5 and the loss tangent ($\tan \delta$) is less than 0.05. Reference markings on the phantom shell allow the complete setup of all predefined phantom positions and measurement grids with respect to the robot.

Figure 6.1: Specific Anthropomorphic Mannequin (SAM) twin phantom manufactured by SPEAG, a plastic container that holds the human phantom tissue used by Kyocera to test the SAR levels emitted by cellphones. The applicator was placed on the flat thoracic section of the manikin.
6.3.2 SAR Set-up and Procedure

The SAR testing procedures included a maximum power test, magnitude of the reflection coefficients (S11) measurements and SAR levels. The first step was to determine the highest input power that would generate a SAR level that is lower than the set SAR limit. Once the maximum input power was determined, magnitude of S11 was measured using a network analyzer (HP 8753E). The magnitude of S11 verified whether proper coupling between the applicator and the phantom shell was established. SAR levels were then measured using the DASY4 system. Operational frequency for both the S11 and the SAR levels was set at 915 MHz. Parameters of the SAR experiment is illustrated in Figure 2.

![Device #_M915 Flat with 15dBm 0mm Air Space, 10-13-09](image)

- **Communication System**: LPD 900, Frequency: 915 MHz, Duty Cycle: 1:1
- **Medium**: M915, Medium parameters used: \( f = 915 \text{ MHz} \); \( \sigma = 1.03 \text{ mho/m} \); \( \varepsilon_r = 54.6 \); \( \rho = 1000 \text{ kg/m}^3 \)
- **Phantom**: SAM 12, Phantom section: Flat Section
- **DASY4 Configuration**: Probe: ET3DV6 - SN1684, ConvF(6.11, 6.11, 6.11), Calibrated: 8/22/2009
- **Sensor-Surface**: 4mm (Mechanical Surface Detection)
- **Electronics**: DAEG Sn603, Calibrated: 9/15/2009
- **Measurement SW**: DASY4, V4.7 Build 71
- **Postprocessing SW**: SEMCAD, V1.8 Build 184
- **Temperature**: Room T = 21.8 +/- 1 deg C, Liquid T = 22.0 +/- 1 deg C

Figure 6.2: Parameters that was utilized for the SAR level testing.

However, since the proposed applicator is designed for human tissue, the applicator would not be matched when it is attached to the plastic shell and result in undesirable reflections of the microwave signal. Consequently, a thin slice
(approximately 1mm thick) of a human equivalent phantom skin tissue was placed between the applicator and the plastic shell for impedance matching. However, in order to verify that the energy of the microwave signal was not dispersed completely into the phantom skin tissue, two applicators for plastic surfaces were also included in the SAR measurements. Placement of the applicator on the bottom flat section of the phantom is illustrated in Figure 6.3. Adhesive tape was used to secure the applicator in place to ensure good contact between applicator and the phantom shell.

In addition, a scaled down model (75% of the original size) of the applicator was also tested to determine the correlation between change in the SAR level and size of the applicator. Therefore, a total of six applicators were tested. Outline of the SAR measurement procedures which is based on the FCC recommended guidelines (FCC, June 2001) are as follows:
Set-up 1: Plastic applicator on the flat section of the phantom shell:

1. Measure output power from the DC power supply with a power meter
2. Place DUT (applicator) on the flat section of the phantom shell
3. Set scan area, grid size, and other setting on the DASY4 software
4. Perform Area Scan, Zoom Scan and SAR Values
5. Verify initial output power with a power meter

Set-up 2: Human equivalent muscle tissue between the human applicator and the flat section of the phantom shell:

1. Measure output power from the DC power supply with a power meter
2. Place human equivalent skin tissue (thickness ≤1 mm) on the flat section of the phantom shell
3. Place DUT (applicator) on the human equivalent skin tissue
4. Set scan area, grid size, and other setting on the DASY4 software
5. Perform Area Scan, Zoom Scan and SAR Values
6. Verify initial output power with a power meter

6.3.3 SAR and S11 Results

Using the setup in Figure 6.3, four input power levels were tested. Although the resulting SAR level at 0.99 W/kg with 50 mW of input power is below the acceptable FCC SAR limit of 1.6 W/kg, an input power of 32 mW was chosen to account for the varying sizes and designs of the applicators.

Table 6-2: Maximum input power level at 15 dBm or 32 mW only results in 0.416 (W/g) which is 2/3 below the allowable FCC SAR level regulations.

<table>
<thead>
<tr>
<th>Input Power</th>
<th>SAR (W/kg) [1g Av ]</th>
</tr>
</thead>
<tbody>
<tr>
<td>(dBm)</td>
<td>(mW)</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td><strong>15</strong></td>
<td><strong>32</strong></td>
</tr>
<tr>
<td>17</td>
<td>50</td>
</tr>
<tr>
<td>20</td>
<td>100</td>
</tr>
</tbody>
</table>
Table 6-2 below consists of the magnitude of S11 and SAR levels for each model. For the magnitude of S11 measurements, all models were below -10 dB, and five of the six applicators were below -20 dB. Therefore it can be concluded that proper coupling between the applicators and the phantom shell was established.

For Devices #3 and #5, the SAR levels were relatively similar at 0.475 W/kg and 0.435 W/kg respectively. This verifies that the thin phantom skin tissue placed between the human applicators (Devices #5, and #6) and the phantom plastic shell did not absorb all of the energy. Therefore, the phantom skin tissue has no or minimal effect on the SAR values because the input energy is coupled to the liquid in the plastic container (SAM). The highest SAR level was measured at 0.475 W/kg with Device #3, which is 1/3 lower than the FCC SAR limit at 1.6 W/kg. The applicators that were scaled down in size to 75% of the original all had SAR levels lower than the original applicators.
Table 6-3: Measured SAR levels from various applicators, designed to couple to plastic and human.

<table>
<thead>
<tr>
<th>Device Applicator Design</th>
<th>Magnitude of S11 (dB)</th>
<th>SAR (W/kg) Limit [1g. Av]</th>
<th>Measured SAR [1g. Av]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1(a)</td>
<td>-22.84</td>
<td>1.6</td>
<td>0.416</td>
</tr>
<tr>
<td>1(b)</td>
<td>-13.15</td>
<td>1.6</td>
<td>0.415</td>
</tr>
<tr>
<td>2(a)</td>
<td>-28.43</td>
<td>1.6</td>
<td>0.475</td>
</tr>
<tr>
<td>2(b)</td>
<td>-26.42</td>
<td>1.6</td>
<td>0.39</td>
</tr>
<tr>
<td>3(a*)</td>
<td>-27.46</td>
<td>1.6</td>
<td>0.435</td>
</tr>
<tr>
<td>3(b*)</td>
<td>-31.52</td>
<td>1.6</td>
<td>0.319</td>
</tr>
</tbody>
</table>
Consequently, the measured applicators are within and more specifically, at least 1/3 lower than the FCC’s SAR safety limit for body worn transmitters. Moreover, the measured SAR levels from these experiments were used in an application to the Institutional Review Board for Human Clinical Studies. The complete application to the IRB is included in Appendix. Preparation for the clinical trials along with Manikin measurements are covered in the next chapter.
Chapter 7 MANIKIN MEASUREMENTS and PREPARATION FOR CLINICAL TRIALS

7.1 Introduction

Preparation for clinical trials using human subjects requires phantom models that are nearly identical to human anatomy, more specifically, human anatomical models for lungs and muscle tissue. The first section of this chapter will cover a comparison between a manual and computer controlled respiration rate using a manikin for medical training at the John A. Burns School of Medicine (JABSOM). This will be followed by an experiment similar to the previous phantom experiments in Chapter 5 using the manikin in place of the thoracic model for fluid accumulation and respiration rate. Lastly, portability and wireless implementation of the microwave system were also investigated.

7.2 Manikin Measurements - Respiration

In this section, two sources of the breathing rate, one from a computer controlled manikin and the other is an airbag manually pumped by a human will be presented. The first subsection will discuss the settings for the computer controlled manikin and manual manikin. Following that will be the results of the phase changes from both of the experiments.

7.2.1 Set-up and Procedure

Similar to the phantom experiments from the previous section, the two applicators are connected to a network analyzer (Agilent PNA E8364B) for the s-parameter measurements. Since the skin of the manikin is made out rubber plastic material the plastic applicator that was previously discussed in Chapter 3 was best suited for the
Two sets of manikins were used for this experiment. The first manikin was connected to a computer that controlled the respiration or breathing rate. The second was intubated which was attached to a _ liter pump. The air pumped into the manikin goes into two rubber air bags to emulate the movement of the lungs during a respiratory cycle. The experimental setup is illustrated in the figures below.

Figure 7.1: Setup of computer controlled and manual breathing manikin experiment.  

Figure 7.2: (Top) Computer controlled breathing rate set at 12 breaths per minute.  
(Bottom) Manual air pump is attached to an intubated manikin to emulate breathing.
Table 7-1: Planar layer model of the setup of the automatic and manual breathing manikin experiment.

<table>
<thead>
<tr>
<th>Material</th>
<th>Thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Applicator</td>
<td>0.066</td>
</tr>
<tr>
<td>Plastic (Chest)</td>
<td>40</td>
</tr>
<tr>
<td>Rubber Bags (Lungs)</td>
<td>15 (deflated)</td>
</tr>
<tr>
<td></td>
<td>50 (inflated)</td>
</tr>
<tr>
<td>Air</td>
<td>75</td>
</tr>
<tr>
<td>Plastic (Back)</td>
<td>5</td>
</tr>
<tr>
<td>Applicator</td>
<td>0.066</td>
</tr>
</tbody>
</table>

For the computer controlled breathing rate, three breathing segments were recorded over a period of about 130 seconds. The first segment had a breathing rate of 12 breaths per minute (BrPM). This was followed by a faster breathing rate of 20 BrPM. After two minutes, the breathing rate was set to 0 BrPM to illustrate the baseline.

For the manual breathing rate with an air bag pump, a breathing rate of 12 BrPM was recorded in addition to the 0 BrPM to establish a baseline in the beginning and ending of the experiment. The recorded phase of the transmission and reflection coefficient of the microwave signal for both experiments will be discussed in the proceeding section.

7.2.2 Results

Phase of the three recorded breathing rates for the computer controlled BrPM are illustrated in Figure 7.3 below. The phase changed for the reflection coefficient (S11) and transmission coefficient (S21) are measured at 1.5° and 2° respectively. Similarly, as seen in Figure 7.3, phase changed from the manual air bag pump was measured at 1° and 3° for S11 and S21 respectively.
Figure 7.3: Phase of the reflection and transmission coefficients of the computer controlled respiration rate experiment

Figure 7.4: Phase of the reflection and transmission coefficients of the manual respiration rat.
Moreover, the computer controlled respiration was further validated using the DSP algorithm that was previously discussed in Chapter 4 to extract the varying breathing rates. The extracted signals are illustrated in Figure 7.5 below. From 0 to 80 seconds, the processed microwave signal indicated a breathing rate of 12 BrPM, followed by 20 BrPM from 80 to 140 seconds which clearly matches the values that were set for the computer controlled breathing rate.

Figure 7.5: Processed reflection and transmission coefficient of the microwave signal from the computer controlled breathing manikin experiment using the proposed DSP.

### 7.3 Manikin Measurements – Respiration and Fluid Accumulation

Similar to the respiration and fluid accumulation experiment in Chapter 5, phase of both the reflection and transmission coefficients were measured. In place of the previous phantom thoracic base model is the manual air bag pump manikin from the previous section. Two applicators were tested for this experiment, the epoxy covered junction and the modified epoxy covered junction. For this chapter, the applicator with epoxy covered junction will be referred to as Ap1 and Ap2 will be the other applicator.
with the modified junction covered by epoxy. The first subsection outlines the materials used for the manikin model and procedures that were conducted. Results of the impedance, magnitude, and phase measurements of the S-parameters for both applicators during the breathing and fluid accumulation will follow. Performance of the two applicators will be compared using the proposed DSP approach from Chapter 4. The section will be concluded with a comparison of the results from the thoracic based model experiments from Chapter 5 and the manikin measurements.

7.3.1 Set-up and Procedure

Similar to the manikin respiration setup from the previous section, the manikin was intubated and attached to a manual airbag pump. Images of the two applicators that were utilized for the breathing and fluid manikin experiments are illustrated in Figure 7.6 below.

![Figure 7.6](image)

Figure 7.6: Two applicators (original with epoxy covered junction – left, and epoxy covered modified junction – right) used in the breathing and fluid accumulation experiment on a manikin.

To emulate pulmonary edema or fluid in the lungs, shredded sponges were added into the rubber air bags. Furthermore, half an inch layer of sponges were placed on top of
the rubber airbags. On the sponge is a phantom muscle tissue to emulate the muscle on
the thoracic region of a person. Planar layer model of the manikin setup as seen on Table
7-2. Fluid was inserted to the airbags and to the layer of sponges using tubes through a 60
cc syringe. As air was pumped into the airbags, the sponges on top of the airbags were
displaced by 2-4 mm, similar to a chest expansion during the breathing cycle. The
applicators were 148 mm apart during inspiration or inhalation and 113 mm during
expiration or exhalation. Illustration of the entire respiration and fluid accumulation setup
is shown in Figure 7.7.

Table 7-2: Planar layer model of the setup of the breathing and fluid accumulation experiment on
a manikin.

<table>
<thead>
<tr>
<th>Planar Layer Model of the Manikin</th>
<th>Thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Plastic (Chest)</td>
<td>40</td>
</tr>
<tr>
<td>Styrofoam</td>
<td>10</td>
</tr>
<tr>
<td>Applicator</td>
<td>0.066</td>
</tr>
<tr>
<td>Muscle</td>
<td>2-4</td>
</tr>
<tr>
<td>Sponge</td>
<td>10</td>
</tr>
<tr>
<td>Rubber Bags (Lungs)</td>
<td>15 (deflated)</td>
</tr>
<tr>
<td>Sponge</td>
<td>75</td>
</tr>
<tr>
<td>Muscle</td>
<td>2-4</td>
</tr>
<tr>
<td>Applicator</td>
<td>0.066</td>
</tr>
<tr>
<td>Styrofoam</td>
<td>10</td>
</tr>
<tr>
<td>Plastic (Back)</td>
<td>5</td>
</tr>
</tbody>
</table>
Figure 7.7: Setup of the manikin experiment that emulated breathing and fluid accumulation which includes an air pump for the airbag filled with sponges and a syringe used to inject fluid into the airbags.

Three sets of experiments were performed. One for the breathing alone, another for just the fluid accumulation and the third one is the combined breathing and fluid accumulation. Three experiments per applicator were conducted for each set of the breathing, fluid accumulation and the combined breathing and fluid accumulation. For the breathing experiment, 15 - 20 seconds of baseline was acquired followed by 1-2 minutes of breathing at a rate of 12 – 15 breaths per minute (BrPM). Similarly for the fluid accumulation, 15 - 20 seconds of baseline was acquired followed by fluid insertion of 2cc every 10 seconds for a total of 30 cc over a period of 150 seconds. Lastly, the experiment with the combined breathing and fluid accumulation is a combination of the previous experiments, which is also similar to the phantom experiments conducted in chapter 5. The experiment was divided into three segments, the baseline, respiration and
fluid accumulation. First, a baseline was established with no external movement for 30 seconds. This was followed by a segment of respiration for about 1 minute. Once a consistent breathing cycle of 12 – 15 BrPM was recorded, fluid was injected to the lungs at a rate of 2cc every 10 seconds. Magnitude and Impedance values were also measured for both the reflection and transmission coefficients to verify if the applicators were matched properly to the phantom muscles. Results of this experiment are included in the following section. Complete protocol and results of this experiment is included in the Appendices.

7.3.2 Results

Measurement of the S-parameters includes the magnitude, phase, and impedance for both of the applicators. Magnitude was measured over the frequency range of 400 MHz to 10 GHz and impedance of each applicator was measured over a span of 200 MHz with center frequency at 915 MHz and 2.4 GHz. The phase was measured during the breathing and fluid accumulation experiments for CW frequencies of 915 MHz and 2.4 GHz. Comparison of the acquired signals from both applicators using the proposed DSP method is also presented.

Magnitude of the reflection coefficients (S11, S22) for both Ap1 and Ap2 were well below -10 dB from 500 MHz – 5 GHz which is plotted in Figure 7.9 below. For Ap1, it was at about -15 dB and -12 dB for 915 MHz and 2.4 GHz respectively. For Ap2, it was measured at about -20 dB and -17 dB for 915 MHz and 2.4 GHz respectively. Moreover, the magnitude of the reflection coefficients for Ap2 was below -15 dB from 500 MHz to 4 GHz compared to Ap1 from 500 MHz to 1.5 GHz. The transmission
coefficients (S21) for both of the applicators were about -45 dB at 915 MHz. However at 2.4 GHz, there was a difference of about -10 dB, -60 dB for Ap2 and -70 dB for Ap1. The measured impedance on the Smith Chart in Figure 7.9 also corresponded with the trends in the reflection and transmission coefficients. The impedance for both applicators at 915 MHz were similar but slightly differed at 2.4 GHz with a difference of 6 – 8 Ω from 50 Ω for Ap1 compared to 3 – 4 Ω for Ap2.

![Magnitude of reflection and transmission coefficients](image)

Figure 7.8: Magnitude of the reflection (top) and transmission (bottom) coefficients of the two applicators used for the manikin experiments. Both applicators have operational frequency of up to 4 or 5 GHz. Ap2 is better matched at the lower frequencies and has a 10 dB gain.
For the breathing and fluid experiment, the changes in the phase of the transmission and reflection coefficients were measured. For the combined breathing and fluid accumulation experiment, the phase changes due to breathing and fluid accumulation were similar to the isolated breathing and isolated fluid experiments. The phase changed in the transmission coefficient due to the fluid accumulation for both applicators were similar at $30^\circ - 40^\circ$ and $75^\circ - 80^\circ$ for 915 MHz and 2.4 GHz respectively. For breathing, it is about $40^\circ$ for both applicators at 915 MHz but Ap2 is higher at 2.4 GHz by $10^\circ$ to $35^\circ$. However, higher noise level is also observed at 2.4 GHz. For fluid accumulation, both applicators were twice as sensitive at 2.4 GHz compared to 915 MHz. For the reflection coefficient, the phase changed from fluid accumulation for both frequencies were about the same for each applicator. However,
Ap2 was more sensitive to the fluid changes by about 3°. Both applicators were more sensitive to the breathing at 915 MHz but Ap2’s phase changes were nearly twice as much of Ap1. S22’s phase changes were less than 1° and clear baseline was not established for all of the experiments. The phase increased or decreased linearly with no clear correlation to the change of water level in the sponges. Summary of the phase changes are included in Table 7-4 and plots of the isolated breathing, isolated fluid accumulation and combined breathing and fluid accumulation are illustrated in Figure 7.12 below.

Table 7-3: Summary of the phase changes due to breathing and fluid accumulation for both applicators. As expected the transmission coefficient is significantly more sensitive to breathing and fluid accumulation compared to the reflection coefficient. At 2.4 GHz, Ap2 is more sensitive to breathing and both applicators are more sensitive to fluid accumulation.

<table>
<thead>
<tr>
<th>FLUID &amp; BREATHING Phase of S11, S22 and S21 for Ap1 and Ap2 at 915 MHz and 2.4 GHz</th>
<th>S11</th>
<th>S21</th>
<th>S22</th>
<th>S11</th>
<th>S21</th>
<th>S22</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ap1</td>
<td>FL</td>
<td>30</td>
<td>2</td>
<td>30</td>
<td>40</td>
<td>5</td>
</tr>
<tr>
<td>915 MHz</td>
<td>BR</td>
<td>30 – 40</td>
<td>2 – 3</td>
<td>&gt; 1</td>
<td>40 – 42</td>
<td>5 – 7</td>
</tr>
<tr>
<td>2.4 GHz</td>
<td>FL</td>
<td>75</td>
<td>1</td>
<td>75</td>
<td>80</td>
<td>5</td>
</tr>
<tr>
<td>BR</td>
<td>40</td>
<td>1 – 1.5</td>
<td>&gt; 0.5</td>
<td>50 – 75</td>
<td>1.5 – 2</td>
<td>&gt; 0.1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>FLUID &amp; BREATHING Phase of S11, S22 and S21 for Ap1 and Ap2 at 915 MHz and 2.4 GHz</th>
<th>Ap1</th>
<th>Ap2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ap1</td>
<td>FL</td>
<td>30</td>
</tr>
<tr>
<td>915 MHz</td>
<td>75</td>
<td></td>
</tr>
<tr>
<td>BR</td>
<td>30 – 40</td>
<td>2 – 3</td>
</tr>
<tr>
<td>2.4 GHz</td>
<td>FL</td>
<td>30 – 40</td>
</tr>
<tr>
<td>BR</td>
<td>40 – 42</td>
<td>5 – 7</td>
</tr>
</tbody>
</table>
Figure 7.10: Phase of the transmission and reflection coefficients at 915 MHz (left) and 2.4 GHz (right) for both of the applicators (Ap1 – blue, Ap2 – red) for the isolated breathing experiment.

Figure 7.11: Phase of the transmission and reflection coefficients at 915 MHz (left) and 2.4 GHz (right) for both of the applicators (Ap1 – blue, Ap2 – red) for the isolated fluid accumulation experiment.
7.3.3 Compare Phantom Muscle and Manikin Experiments using DSP

Using the proposed DSP, results of the breathing and fluid accumulation experiments can be further quantified. Since the results are nearly identical for the three experiments, only one set of data per applicator at 915 MHz will be processed using the proposed DSP. Furthermore, processed data were compared to the phantom experiments conducted in Chapter 5. Results of the proposed DSP processed reflection and transmission coefficients are illustrated in the figure below, which includes Applicators 1 and 2 for the manikin experiments and the original applicator for the phantom experiments.
Figure 7.13: Amplitude and rate of the respiration rate in addition to the fluid accumulation were extracted from the transmission (right) and reflection (left) coefficient of the microwave signal from the manikin and phantom experiments at 915 MHz using the proposed DSP algorithm.
Amplitude of the phase change due to the breathing and fluid accumulation will be the main parameter that this section is based on. Although the trend is the same, phase changes in the transmission coefficient (S21) are more significant compared to the reflection coefficient (S11) for both breathing and fluid accumulation for all of the applicators. However, the distance between the two applicators (original) for the phantom experiment was a lot smaller at 50 mm compared to 120 mm for the manikin experiment. From this, it can be concluded that given the same distance between the applicators for manikin and phantom experiments, the original applicators may not show the same amount of sensitivity for the transmission coefficient. All of the applicators indicated greater sensitivity to the fluid changes compared to breathing. Ap2 was the most sensitive of the three applicators for fluid accumulation changes by 10° - 15° for S21 and the original applicator showed comparable sensitivity of 5° for S11.

Lastly, since the phase changes for the transmission and reflection coefficients from all of the applicators were easily quantifiable after it was processed with the proposed DSP even for less than 1° phase change, it can be concluded that its ability to extract the breathing and fluid accumulation from the microwave signal has been successfully validated. Additional preparation for clinical implementation of the microwave system will be discussed in the proceeding sections.

7.4 Preparation for Clinical Implementation

The existing measurement setup included a Network Analyzer which made it difficult and impractical for continuous, ambulatory, bedside monitoring device. Consequently, efforts were made to build a smaller and lighter, low cost single frequency
receiver. In addition, wireless implementation using Bluetooth was investigated. More specifically, measurements were conducted to quantify the threshold of the frequency offset from the operating frequencies of 915 MHz and 2.4 GHz that will cause interference to the microwave signal.

### 7.4.1 Low Cost Receiver Design

To simplify the design of the receiver, a single frequency receiver was built using a voltage controlled oscillator, low noise amplifiers (LNA) and a mixer which was powered by eight AA batteries. The receiver was connected to a BNC connector that led to the National Instruments Data Acquisition (NI - DAQ) unit. The DAQ measured the voltage changes and converted it into a discrete signal that was sent to the laptop through a USB cable. The LabView GUI was used to acquire the measured voltage and displayed the corresponding phase. Voltage equivalence of the phase change from breathing was first determined by connecting the applicators to a network analyzer. Changes due to breathing were approximately measured at 7° phase change which is equivalent to 0.25 V for the NI – DAQ. Block diagram of the network analyzer replacement unit and the low cost receiver is illustrated in the figure below.
Figure 7.14: Block diagram of the network analyzer replacement system and the low cost receiver (3x5 inches). The system consists of the low cost receiver, NI – DAQ (for data acquisition), and processing unit with the LabView GUI.
7.4.2 Low Cost Receiver Discussion and Results

Similar to the phantom experiments, breathing and fluid accumulation experiments were conducted. On the LabView GUI, the channel was set to acquire data from the NI – DAQ. The raw data from the DAQ was multiplied by 7° and stored as the phase of the signal. The figure below contains the acquired raw transmission coefficient from the NI – DAQ and the corresponding calculated phase. With the proper scaling, it can be concluded that the measured voltage and the calculated phase change is nearly identical and well matched. Although the system is limited with just a single frequency at 915 MHz, it is a lot smaller and more compact (3x5 inches) compared to a network analyzer. The receiver system can be further optimized by adding more amplifiers and replacing the power supply with a more efficient and smaller source of power. Since the NI – DAQ unit is currently connected to the laptop through a USB cable; the unit may be replaced with another DAQ that has wireless capabilities to transmit the data wirelessly. However, other factors such as RF interference from other devices as well as its effects on the microwave applicator will need to be considered.
Figure 7.15: Measured voltage from the NI – DAQ unit that was acquired during a breathing and fluid accumulation phantom experiment. The voltage was multiplied with the corresponding phase change initially measured on a network analyzer. A 0.25 V was equivalent to a 7° phase change.

7.4.3 Wireless Implementation Set-up and Procedure

The interference measurements at 915 MHz and 2.4 GHz were conducted in an anechoic chamber. The transmitter (horn antenna) was 20 feet away from the applicators (modified epoxy covered junction applicator – app2 from the manikin experiments). The horn antenna was oriented for both vertical and horizontal polarization 7.7 dB Gain at 915 MHz and 10.6 dB Gain at 2.4 GHz. Due to the distance (20 ft.) between the applicators and the horn antenna, the worst case scenario of an input power of 360 mW was used. A network analyzer (Agilent PNA E8364B) was used to measure the transmission and reflection coefficients of the applicators with an IF Bandwidth of 35 kHz. Setup of the applicators is similar to the lung phantom experiments in Chapter 5. Setup of the interference experiment is illustrated in Figure 7.16.
Prior to the interference experiment, the impedance and magnitude of the reflection and transmission coefficients of the applicators were measured. This was then followed by two sets of experiments, interference at 915 MHz and 2.4 GHz respectively. Similar to the human phantom lung experiment, interference for each experiment was quantified by the change in the phase of the transmission and reflection coefficients. For the first set of experiments, the operational frequency for the applicators was set at 915 MHz. For each polarization, the horn antenna was set to five incrementing frequency offsets of 0.05 MHz from 915 MHz, which includes 915.01 MHz to 915.2 MHz. Similarly, the 10 experiments were repeated with the same incremental frequency offset for the interference measurement at 2.4 GHz, from 2.40001 GHz to 2.4002 GHz. A total of 20 measurements were conducted for the frequency interference measurement.

Figure 7.16: Setup of the RF Interference from similar devices at 915 MHz and Bluetooth (2.4 GHz) devices in an anechoic chamber with a horn antenna as the source of the RF interference.
7.4.4 Wireless Implementation Results and Observations

The measured impedance and magnitude of the applicators will be presented first and followed by graphs of the interference offset for 915 MHz and 2.4 GHz. Magnitude of S11, S22, and S21 from 400 MHz to 10 GHz are illustrated in Figure 7.18.

![Graph](image.png)

Figure 7.17: Magnitude of the transmission and reflection coefficients for the epoxy covered modified junction when coupled to the phantom muscle for the RF interference experiment.

As the operational frequency of the transmitter nears the frequency of the applicators, interference to the reflection and transmission coefficients of the applicators increases. Horizontal polarization of the transmitter also causes greater disturbance to the signal between the applicators compared to vertical polarization. Threshold of allowable frequency offset .15 and .05 MHz for 915 MHz and 2.4 GHz respectively. This means that devices emitting RF frequencies .15 MHz more or less than 915 MHz and .05 MHz more or less than 2.4 GHz will interfere with the microwave signal of the applicators. For
most cases, horizontal polarization of the transmitter causes greater interference compared to vertical polarization. When the transmitter is operating at the threshold frequency of 915.15 MHz with horizontal polarization, 5°, 4°, and 1° phase changed is observed for S21, S22 and S11 respectively. Similarly, at the frequency threshold of 2.40005 GHz, degree phase changed of S21 is observed at 800° and 1500° for horizontal and vertical polarization respectively. Although the frequency offset is .10 MHz better for the operating frequency of 2.4 GHz compared to 915 MHz, the change in the phase of the signal caused by the interference is far greater at 800° (at 0.5 MHz offset for 2.4 GHz) as opposed to the 5° (at 0.15 MHz offset for 915 MHz) for S21. At the threshold frequency offset for both 915 MHz and 2.4 GHz, the changes in the phase for the reflection coefficients are similar and ranges from 1° to 5°. Table 7-4 summarizes the phase change of the microwave signal of the applicators from the interference caused by the horn antenna which is also plotted in Figure 7.18. Furthermore, sample of the acquired signal at the frequency threshold offset for both operational frequencies are also included in Figure 7.18.
Figure 7.18: Changes in the phase of the transmission coefficient as the frequency of the RF interference (from a horizontally polarized horn antenna) nears the operational frequency of the microwave signal at 915 MHz (left) and 2.4 GHz (right).

Figure 7.19: Graphical representation of the RF interference of the microwave signal from horizontally and vertically polarized horn antenna at 915 MHz and 2.4 GHz.
Table 7-4: Summary of the RF interference threshold by varying the frequency offset when the operational frequency is set to 915 MHz and 2.4 GHz.

<table>
<thead>
<tr>
<th>TX Frequency at 915 MHz +/- (offset MHz)</th>
<th>ΔPhase (Degrees)</th>
<th>TX Frequency at 2.4 GHz +/- (offset MHz)</th>
<th>ΔPhase (Degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Horizontal Pol S21 S11 S22</td>
<td>Vertical Pol S21 S11 S22</td>
<td>Horizontal Pol S21 S11 S22</td>
</tr>
<tr>
<td>.01</td>
<td>10k 75 10k</td>
<td>25 12 12</td>
<td>.01</td>
</tr>
<tr>
<td>.05</td>
<td>50 10 30</td>
<td>5 1.5 3</td>
<td>.05</td>
</tr>
<tr>
<td>.10</td>
<td>10 1 5</td>
<td>0 0 0</td>
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</tr>
<tr>
<td>.15</td>
<td>5 1 4</td>
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</tr>
<tr>
<td>.20</td>
<td>0 0 0</td>
<td>0 0 0</td>
<td>.20</td>
</tr>
</tbody>
</table>
Chapter 8 CONCLUSIONS AND FUTURE WORK

8.1 Summary and Conclusions

The prevalence and costs of heart related diseases such as HF requires immediate long term solution as we face the imminent shortage of health care providers and aging population. Government entities such as NIH, has urged the need for innovation in the area of Telemedicine or e-medicine, for early detection and monitor cases of HF.

This study validated the viability of implementing and developing a noninvasive ambulatory microwave based technology applicator in a HCAC’s existing interactive telemedicine system for hemodynamic assessment as an alternative solution to reducing the costs, morbidity and mortality rate from heart related diseases such as HF. Feasibility to detect early cases and symptoms of HF by monitoring fluid content and respiration rate as well as heart rate has been proven through the phantom and manikin experiments. These experiments utilized the proposed DSP vital sign extraction algorithm that separated the signals acquired by the microwave applicator. Furthermore, SAR levels for RF safety regulations and RF interference experiments were also conducted to address challenges for practical implementation in clinical environment. Results from these experiments were submitted in an IRB proposal to conduct human clinical studies which was approved by April 2011. Summary of these experiments is outlined in the proceeding paragraphs.

Based on the pre-existing applicator designs, prototypes of various applicators were fabricated for RF safety and human hemodynamic assessment clinical trials. Furthermore, the proposed DSP algorithm based on STFT and linear regression,
accurately extracted multiple vital signs such as the heart rate and respiration rate in addition to the fluid accumulation. All of which are critical to patients who suffers from HF and pulmonary edema.

In Chapter 1, advantages and limitations of existing invasive and noninvasive devices used for hemodynamic assessment was discussed. Noninvasive devices are organized into four broad categories of transduction; External chest mechanical signals; External chest electrical signals; and Ultrasound techniques. One in particular, is the electrical impedance cardiography method (ICG), which was compared to the measurement technique of the microwave applicator in Chapter 2. Since microwave signals have greater penetration depth, it provides more resolution to the changes in the electrical properties of vital organs such as the heart and the lungs. Prior investigations on animal lung experiments, Iskander et. al concluded that 915 MHz was the optimum operational frequency due to the attenuation and resolution trade off. Higher frequencies provided more resolution but decreased in depth penetration.

Numerical analysis on HFSS and fabricated prototypes of the microwave applicator was included in Chapter 3. Various applicator designs were developed to improve the bandwidth, resolution and coupling capabilities to other materials such as plastic which was later used for the RF safety testing in Chapter 6. Six applicators models were presented which included the original applicator with the epoxy, modified junction, broadband applicator, smaller structure, and two plastic applicators. The original applicator with the epoxy on the junction allowed better transition of the energy to the rest of the structure rather having all the energy coupled to the muscle just at the junction.
The design of the modified junction applicator accounted to match the epoxy material at
the junction rather than to muscle. Width of the center conductor and the gap between the
ground plane and the center conductor was also adjusted to increase the bandwidth of the
applicator. Results from the numerical analysis on HFSS indicated of an improvement of
-5 dB in magnitude of S11 from 1 GHz to 2.4 GHz and extended to 7 GHz with
magnitudes lower than -10 dB. A smaller applicator was also designed to provide better
coupling to uneven surfaces on the thoracic region. Although smaller, magnitude of S11
was still less than -15 dB for frequencies less than 2 GHz. To prepare for clinical trials,
RF safety experiments were also conducted which included plastic containers that held
the phantom tissue that was used to determine the amount of Specific Absorption Rate.
Consequently, applicators designed to couple to plastic material was also created.

In Chapter 4, proposed signal extraction algorithm based on STFT and linear
regression was utilized to extract vital signs and fluid accumulation in the lungs from the
microwave signal was presented. The algorithm was initially validated using the heart
signal from previous animal experiments. It’s accuracy to separate or extract the signals
were further confirmed in the phantom and manikin experiments in Chapter 5 and
Chapter 7 respectively. With the proposed DSP, trend of the phase of the transmission
and reflection coefficient of the microwave signal during water accumulation clearly
follows the trend of the raw data. Rate and amplitude of the breathing was extracted at
0.08 Hz and 7° – 10° respectively for the transmission coefficient. Experiments were also
conducted that verified the dielectric property of the phantom human tissues that was
used in the lung phantom thoracic based model that emulated respiration and pulmonary edema or fluid in the lungs.

With the promising results from the phantom experiments and in preparation for the clinical trials, Chapter 6 discussed the RF Safety experiments that were conducted in Kyocera Wireless labs (San Diego, CA). The DASY 4 system (SPEAG) indicated that the Specific Absorption Rate at 32 mW input power was 1/3 below the FCC standard for maximum allowable SAR level. Applicators used in the experiment included the applicators designed for the plastic container that held phantom muscle material. Measured SAR levels of plastic applicators were similar to the original applicator designed for human thoracic muscle. This ensured that the power from the original applicator was not dispersed into the thin layer (< 1 mm) of phantom muscle that was used for matching.

After the FCC safety levels were met, Chapter 7 presented the experimental trials that utilized life-sized human thoracic manikins provided by the medical group at Telehealth Research Institute (TRI) at the John A. Burns School of Medicine (JABSOM). Breathing was emulated using a computer controlled manikin which is similar to the breathing cycle that was manually conducted on the thoracic based phantom model in Chapter 5. Layer of sponges were added to the manikin model to account for the water accumulation during the breathing cycle. The original applicator and modified junction applicator clearly detected the respiration rate and fluid accumulation. Of the two applicators, the modified junction showed greater sensitivity for both S11 and S21. Signals were successfully extracted using the proposed DSP algorithm. Moreover, the
interference experiment conducted in the anechoic chamber revealed that interference occurs when a signal is within 0.15 MHz of 915 MHz and 0.05 MHz and 2.4 GHz.

Results from these experiments were used in the proposal for the Institutional Review Board for human clinical trials. The proposal was officially approved and it will be in collaboration with the Tele Health Research Institute at JABSOM.

8.2 Future Work

For future work, there are three main areas of study which includes; 1) applicator and system design, 2) DSP algorithm, and 3) clinical trials. For the applicator, additional designs are needed to make it more broadband. Flexibility with the frequency would allow more applications in other areas such as imaging and glucose monitoring. Further studies on the tradeoff between sensitivity and attenuation at other frequencies may eliminate the need for two applicators to measure the transmission coefficient. If one applicator would provide quantifiable data from the phase change in the reflection coefficient, the challenge with placing the applicators in line of each other would be resolved. Furthermore, measurements are currently made using network analyzer which is cumbersome and hinders the mobility of the applicator. Therefore more development is needed to integrate the applicator with a smaller and simpler network analyzer even if it’s just for a single frequency. With focus on mobility and ease of the applicator, another consideration is to develop a wireless system data acquisition system. This will also eliminate the clutter of wires and noise from the cables. However, moving towards a wireless system would require a more complex system for data acquisition. This would
require a more robust DSP algorithm not just for signal extraction but for noise and signal filtering as well. Consequently, more human clinical trials need to be conducted not only to determine the sensitivity to human vital signs but also investigate other factors in a clinical environment that may significantly affect the applicator’s performance.
Appendix 1  RF SAFETY

See Attached Document
MEMORANDUM

April 19, 2011

TO: Magdy F. Iskander, Ph.D.
Principal Investigator
College of Engineering

FROM: Nancy R. King
Director

SUBJECT: CHS # 18228, "Non-Invasive Microwave Measurements of Human Cardio-Respiratory Hemodynamics"

This is to acknowledge receipt of your email response received March 31, 2011, to the stipulations issued by the Committee on Human Studies (CHS) during its review of the project identified above at its meeting on March 16, 2011. The information you provided satisfactorily addressed CHS stipulations, and the project, is approved for one year effective April 18, 2011.

This memorandum is your record of CHS approval of this study. Please maintain it with your study records.

CHS approval for this project will expire on April 17, 2012. If you expect your project to continue beyond this date, you must submit an application for renewal of this CHS approval. CHS approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes, you must obtain CHS approval prior to implementing them. Unanticipated problems that are likely to affect study participants must be promptly reported to the CHS.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by CHS and other authorized agencies.

Please notify this office when your project is completed. Upon notification, we will close our files pertaining to your project. Reactivation of CHS approval will require a new CHS application.

Please contact this office if you have any questions or require assistance. We appreciate your cooperation, and wish you success with your research.
## Appendix 2  Protocol for the Manikin Experiment

<table>
<thead>
<tr>
<th>Applicators ($A_n$)</th>
<th>Frequencies ($F_N$):</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Original with Epoxy on junction</td>
<td>1. 915 MHz: 900 MHz – 930 MHz</td>
</tr>
<tr>
<td>2. Modified junction covered with Epoxy</td>
<td>2. 2.4 GHz: 2.385 GHz – 2.415 GHz</td>
</tr>
<tr>
<td>3. Plastic Applicator</td>
<td>3. 400 MHz – 10 GHz</td>
</tr>
</tbody>
</table>

### Protocol for each applicator

#### I. CALIBRATION: Perform and store 2-port calibration on PNA E83645B

1. Calibration1: Impedance and Magnitude Measurements
   a. Number of points: 1601
   b. IF Bandwidth: 1kHz
   c. Sweep Type: Linear Frequency
   d. Frequency Range: 400 MHz – 10 GHz

2. Calibration2: Phase Measurements
   a. Number of points: 101
   b. IF Bandwidth: 1kHz
   c. Sweep Type: Continuous Wave (CW)
   d. Frequency: 915 MHz, 2.4 GHz

#### II. IMPEDANCE and MAGNITUDE Measurements

1. Initialize Calibration1
2. Set Frequency Range1 (900 MHz – 930 MHz)
3. Measure Impedance for S11 and S22
4. Measure Magnitude for S11, S22 and S21
5. Repeat Step II.2-II.4 for 2.385 GHz – 2.415 GHz
6. Repeat Step II.2-II.4 for 400 MHz – 10 GHz

#### III. NOISE Measurements

1. Magnitude
   a. Acquire and store Magnitude trace of S11
   b. Subtract Data from the memory trace stored from a.
   c. Touch and move cable
   d. Store the data displayed on network analyzer. This is the difference between first stored data and the data from the cable movement. This data is the noise from motion artifacts from the cable.
   e. Repeat noise measurement procedure for S22
   f. Repeat Step II. And Step III. for Frequency Range2 (2.385 GHz – 2.415 GHz)

2. Phase
   a. Initialize Calibration2 for 915 MHz
   b. Acquire s-parameters using LabView program
      i. Develop a baseline for a minimum of 30 seconds by not performing any tasks on the manikin.
      ii. Place hand 4 inches above the applicator on the chest for 30 seconds.
      iii. Move hands away from the applicator and wait for 30 seconds. Then touch the coaxial cable for the applicators one at a time for 30 seconds.
   c. Repeat Step III.2. for 2.4 GHz

#### IV. BREATHING Measurements
1. Initialize Calibration2 for 915 MHz
2. Acquire S-Parameters using LabView program
   i. Develop a baseline for a minimum of 30 seconds by not performing any tasks on the manikin.
   ii. Pump the manual air pump for at least two minutes at a rate of 12-15 Beats Per Minute (BPM).
3. Repeat IV two more times for a total of three breathing measurements at 915 MHz
4. Repeat Step IV for 2.4 GHz

**V. FLUID Measurements**

1. Measure the weight (dry weight) of the lung bags and the sponge lungs.
2. Initialize Calibration2 for 915 MHz
3. Acquire S-Parameters using LabView program
   i. Develop a baseline for a minimum of 30 seconds by not performing any tasks on the manikin.
   ii. Insert the 30 cc fluid into the tube at a rate of 2cc per 10 seconds.
4. Measure the weight (wet weight) of the lung bags and the sponge lungs.
5. Repeat V two more times for a total of three fluid measurements at 915 MHz
6. Repeat Step V for 2.4 GHz
REFERENCES


