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Frequency Dependent Planar Electromagnetic Modeling of Human Body and Theoretical Study on Attenuation for Power Budget Estimation of UWB Radar

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Frequency Dependent Planar Electromagnetic Modeling of Human Body and Theoretical Study on Attenuation for Power Budget Estimation of UWB Radar

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I. Introduction

For many applications such as lie detection, life detection etc. remote detection/monitoring of cardio-pulmonary features of a human being such as heartbeat rate, breath rate, and blood flow rate etc. are necessary. This can be done to good accuracy using high resolution radars like ultra wide-band radars. Moreover, human body monitoring using UWB technology can permit the subject to be in absolutely free space as it uses radiated field unlike other methods e.g. functional-MRI (f-MRI) uses induced EM field. A summary of the existing models found in the literature are presented below.

Radar monitoring of human physiologic functions started in early 70's as mentioned in [1] but further progress was stopped as the technique was expensive. Micro-power Impulse Radar (MIR), an UWB radar model was developed by Thomas E. McEwan at Lawrence Livermore National Laboratory (LLNL) during 1993 which was able to detect the movements of heart wall from the difference of reflection magnitude between various tissue interfaces. This model proposed in both of the patents on medical UWB radar awarded to McEwan [2-3], did not account for the various living tissues through which the UWB pulse travels before it strikes the heart wall and hence is quite ingenuous. Another more accurate model described in [4] was found at Tor Vergata University of Rome, Italy that accounted for thickness, impedance, linear attenuation and wave velocity of six superimposed living tissues (air, fat, muscle, cartilage, deflated lung and heart) encountered by the UWB pulse passing through the chest from skin to heart. This model was based on the data taken from Visible Human Project [5] and Gabriel's data book of dielectric properties of tissues [6]. As an immediate application of this model the author of this model [4] developed the UWB radar based stealthy lie detector [7]. The attenuation predicted by the model given in [4] was linear but the imaginary part of reflection coefficient and multiple reflections were ignored. In [4], author himself mentioned that this new model remains intrinsically wrong as the dielectric properties used were measured on living tissues using a continuous-wave (CW) of frequency 1500 MHz. The analysis was based on only one frequency and therefore was frequency independent. Hence the author suggested for a more effective model by using ultra wide-band dielectric properties instead of narrow band ones for which a convolution method or a Finite Difference Time Domain (FDTD) technique should be used and also by considering both the real and imaginary parts of the reflection coefficients at the boundaries because the UWB receiving correlator, working in the time domain, is strongly sensitive to phase errors. In 2006, two frequency-dependent EM modeling techniques: planar

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technique and FDTD technique were presented [8] which could predict the response of a four layer biological model with three layers of normal tissue and one layer of cancerous soft tissue to input UWB signal. The thickness of each layer in the models was taken from the Visible Human Project [5] and dielectric properties of tissues from Gabriel's data book [6] as followed by the earlier model presented in [4]. As far as the planar technique is concerned the authors mentioned that the technique produces reasonable results for applications of less number of layers but not very accurate for more complex layers as it does not consider to model multiple reflections and hence suggested that the planar technique could be extended to include provision for multiple reflections.

A circuital model for monitoring of cardio-pulmonary activity with UWB radar of operating frequency range of 0.1 to 3 GHz was presented in 2008 by Pisa et.al [9]. This model was developed taking the signal source, transmitting and receiving antenna properties as well as a planar model of human thorax into account. The thorax model considered the same tissues: air, fat, muscle, cartilage, lung and heart as used in the model employed in [4]. The attenuation of the air-heart-air path was evaluated over the ultra wideband frequency of 0.1 to 3 GHz for two types of lungs: deflated lung and inflated lung. It was found that the model has predicted attenuation increase with frequency and moreover the amount of attenuation increase is more in case of deflated lung than that in case of inflated lung [9]. This planar model was a circuital model but, an electromagnetic (EM) model could be a more approximate model over circuital model.

Both the UWB models reported in [4] and [9] ignored the skin tissue in the thoracic modeling but attenuation due to skin tissue is also significant because skin is also one of the greatest electromagnetic field absorbing tissues like muscle and blood in the human body as reported in [10]. Moreover, the circuital model given in [9] is frequency dependent whereas the model described in [4] was developed for a continuous wave of 1.5 GHz and hence was a frequency independent model. A frequency dependent planar model described in [8] considers skin but did not include the phenomenon of multiple reflections.

Also, calculations pertaining to incident, reflection and transmission components of power at every interface are considered, which are not calculated and reported by any of the models. An investigation of these effects such as accounting for the electromagnetic information due to skin, the phenomenon of multiple reflections, frequency dependence of signal attenuation is the subject of this paper.

In this paper, two frequency-dependent planar electromagnetic models of human thorax: the intrinsic impedance or classical model and the impedance transformation model are reported in which both models incorporate the dispersive behavior of selected biological tissues including skin. When the impedance transformation model takes care of the effect of multiple reflections, the classical model does not consider that effect. First, without including skin in the modeling, the signal attenuation results are obtained and compared with the attenuation results reported in the circuital model of [9]. Then the signal attenuation results from the heart with skin included in both the models are analyzed and compared.

As skin tissue is included in the thoracic modeling and particularly, the multiple reflections are also taken care of by the impedance transformation model the signal attenuation results predicted by the models presented in this paper are considered to be more realistic than the model-predicted attenuation of the circuital model reported in [9].

II. Theoretical Background

In order to study the behavior of the backscattered field from a human body illuminated by plane electromagnetic waves of a radar transmitter, simplification of the problem is considered by modeling the human body as a series of biological tissue layers of complex impedances. Knowing the dielectric properties of the biological tissues, and by utilizing the basic principles of electromagnetic wave propagation in accordance with the physical processes that determine the speed of propagation and the amount of attenuation, the power received by the radar receiver, the electric field (E-field) reflection coefficients at every interface and power reflection coefficient of the heart can be determined.

a) Electromagnetic Modeling of Human Body

In order to model the electromagnetic response of human body to radar waves, it is necessary to know with sufficient precision the electromagnetic characteristics of typical biological tissues used to build the body. For this purpose, the transverse section of human anatomy taken from web [11] as shown in Fig.1 is referred. The figure depicts the human thorax showing the contents of the middle and the posterior mediastinum. The pleural and pericardial cavities are exaggerated since normally there is no space between parietal and visceral pleura and between pericardium and heart. Various tissues like sternum, hard cholesterol, muscle, ribs, costal pleura, pleural cavity, pulmonary pleura, pericardium cavity, heart, pulmonary artery (left and right), aorta, internal mammary vessels, thoracic aorta, esophagus, bronchus, thoracic duct, body of thoracic vertebrae, lungs (left and right) etc. encounter in the path of propagation partly or fully, when the human body is illuminated by the radar wave. Let us consider only some of the major biological tissues such as skin, fat, muscle, cartilage, lungs (deflated and inflated separately) and heart as the fundamental layers for a simple one-dimensional modeling of human body. As electromagnetic behavior due to heartbeat movements are of interest, assuming the radiation exposure from anterior of the of the human body to the posterior through heart, the tissue structure considered in view of the electromagnetic body modeling is as shown in Fig.2.

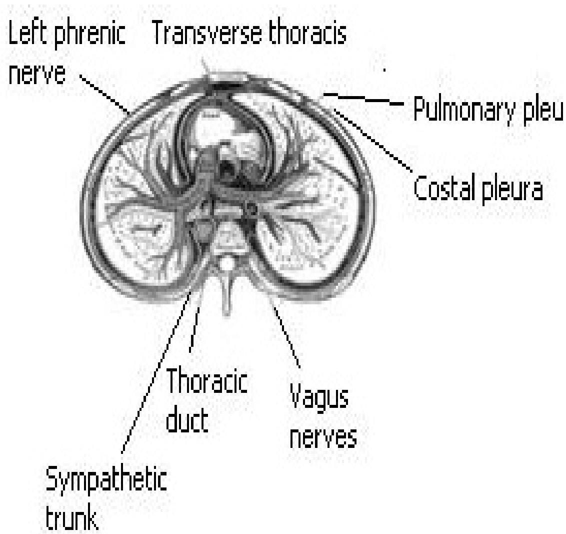
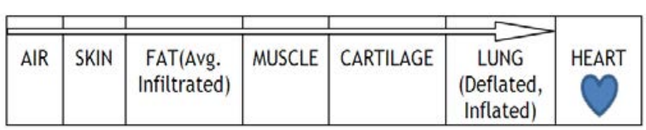
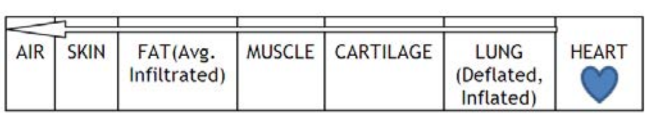


Figure 1 : Transverse section of human thorax [11]



(a)



(b)

Figure 2 : Tissue structure for EM modeling of human body (a) forward propagation (b) backward propagation

The anatomical thickness values of the tissue layers (Table 1) considered in both of the models are based on the Visible Human Project [5] like those already employed in the UWB models of [4], [7-10].

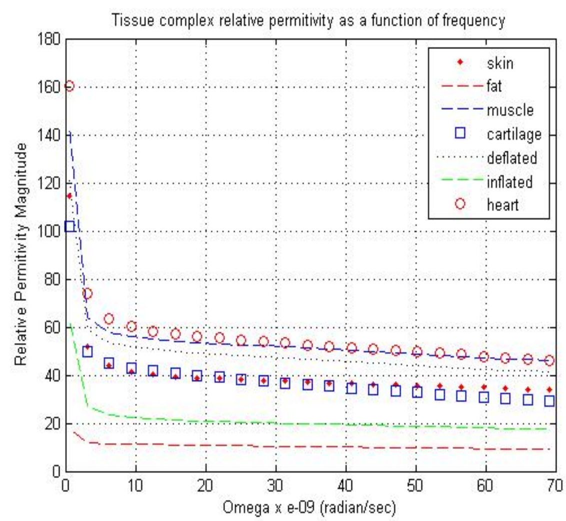
Table 1 : Anatomical thickness of tissue layers

Tissue name	Thickness (mm)
Air	200
Skin(Dry)	1.5
Fat	9.6
Muscle	13.5
Cartilage	11.6
Lung	5.78

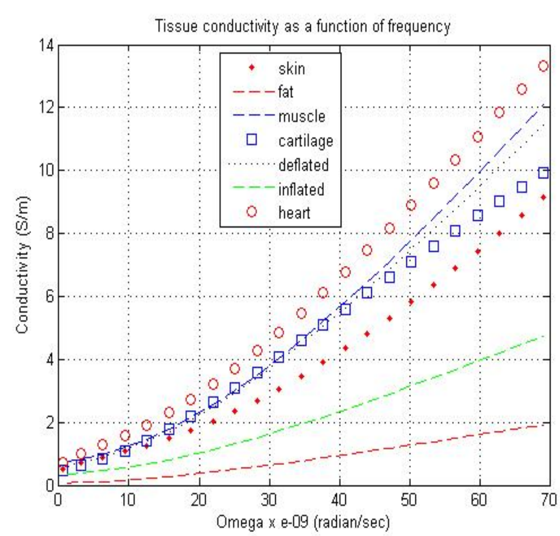
b) Frequency dependent dielectric properties of tissues in human thorax

Based on the Cole and Cole equation as mentioned in [9] that describes the dispersive behavior of tissues, the frequency dependent dielectric bodies of human body tissues are computed as given in the Gabriel's data book of dielectric properties of tissues [6] and are also reported on the web [13].

In our work, the dielectric properties such as complex relative permittivity, conductivity, attenuation constant, phase-shift constant, complex impedance and velocity of wave propagation of the selected tissues for modeling at all frequencies of 0.1 to 3 GHz ultra-wideband are taken from [6]. The variations of the above dielectric properties as a function of frequency are plotted in Fig.3. It is seen that as frequency increases, relative permittivity decreases, but conductivity, attenuation constant, phase-shift constant and intrinsic impedance increase as shown in Fig. 3(a) – (e) respectively.



(a)



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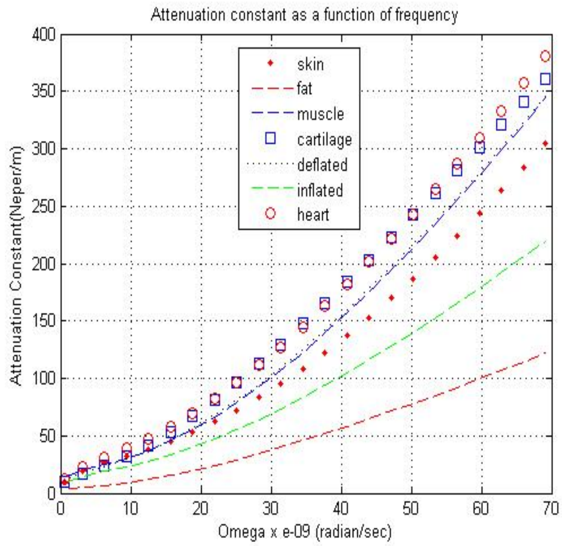

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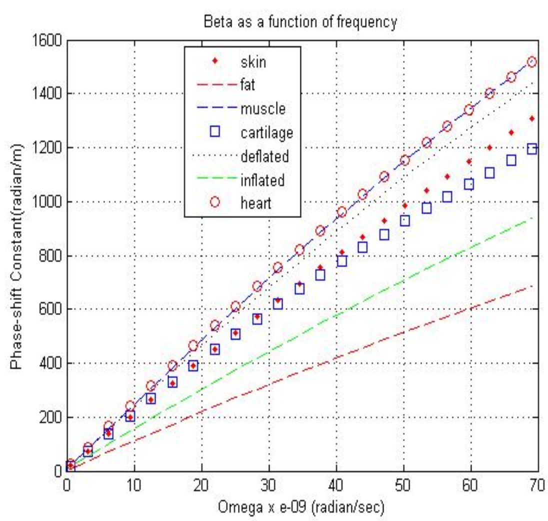

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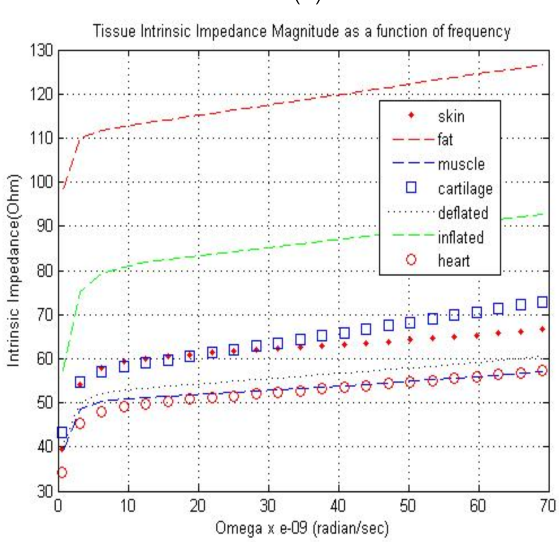


(e)

Figure 3 : Variation of (a) relative permittivity (b) conductivity (c) attenuation constant (d) phase-shift constant (e) impedance of tissues with frequency

III. Electromagnetic Analysis

When an electromagnetic wave is incident on an interface between two consecutive tissue layers a part of the EM energy is transmitted and the remaining is reflected [8] as shown below in Fig.4. The amount of reflection and transmission of energy depends on the frequency dependent dielectric properties of the layers on either side of the interface.

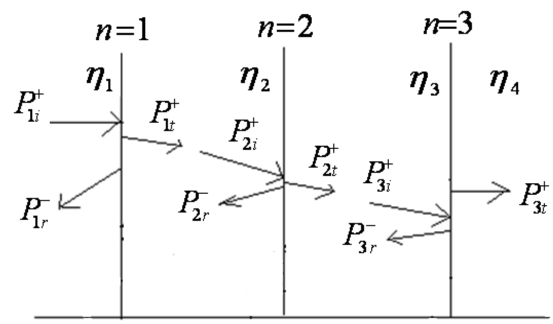


Figure 4 : Intrinsic Impedance Model

As biological tissues are lossy media, from an electromagnetic point of view they are characterized in terms of the attenuation constant, phase-shift constant, conductivity, complex permittivity, complex impedance, complex reflection coefficient etc. These characterization parameters are highly frequency dependent. Therefore, the reflected and transmitted power at all tissue interfaces calculated at every individual frequency of the ultra wide-band range using the concept of EM wave propagation in lossy media are as described in [14].

a) Propagation in Lossy Media

Wave losses due to electric field are described through complex permittivity, ϵ. Similarly, complex permeability, μ is used to model the losses owing to medium response to the magnetic field.

But from wave propagation point of view in most of the materials, the magnetic response is usually very weak compared to the dielectric response. Therefore,

$$\mu \cong \mu_0 \text{ i.e. } \mu_r^{'}=1, \mu_r^{''}=0$$

Consequently, wave losses can be confined to the loss due to the complex permittivity only assuming μ entirely real for the analytical loss calculation.

As in [14-15], for a lossy dielectric media

$$D = \epsilon E \quad (1)$$

$$J = \sigma E \quad (2)$$

$$B = \mu H \quad (3)$$

where

$$\epsilon\text{,Complex permittivity} = \epsilon' - j\epsilon'' = \epsilon_0\left(\epsilon_r' - j\epsilon_r''\right) \quad (4)$$

and

$$\mu\text{,Complex permeability} = \mu' - j\mu'' = \mu_0(\mu_r' - j\mu_r'') \quad (5)$$

b) Characterization of loss

Considering the human tissues as homogeneous, isotropic, and dispersive media, the characterization parameters such as complex propagation constant (γ), complex attenuation constant (α), complex phase-shift constant (β) and complex intrinsic impedance(η) as defined in [14-15] are given as below.

Propagation constant, $\gamma = \alpha + j\beta = \omega\sqrt{\mu\epsilon} = \omega\sqrt{\mu(\epsilon' - j\epsilon'')}$

$$= \sqrt{\mu\epsilon'\left(1 - j\frac{\epsilon''}{\epsilon'}\right)} \quad (6)$$

Now,

$$\text{Re}\{\gamma\} = \alpha\text{, attenuation constant}$$

$$= \omega\sqrt{\frac{\mu\epsilon'}{2}}\left(\sqrt{1+\left(\frac{\epsilon''}{\epsilon'}\right)^2} - 1\right)^{1/2}$$

$$= \omega\sqrt{\frac{\mu\epsilon'}{2}}\left(\sqrt{1+(\text{LossTangent})^2} - 1\right)^{1/2} \quad (7)$$

and

$$\text{Im}\{\gamma\} = \beta\text{, phaseshift constant}$$

$$= \omega\sqrt{\frac{\mu\epsilon'}{2}}\left(\sqrt{1+\left(\frac{\epsilon''}{\epsilon'}\right)^2} + 1\right)^{1/2}$$

$$= \omega\sqrt{\frac{\mu\epsilon'}{2}}\left(\sqrt{1+(\text{LossTangent})^2} - 1\right)^{1/2} \quad (8)$$

where $\quad \text{LossTangent} = \frac{\epsilon''}{\epsilon'} = \frac{\sigma}{\omega\epsilon'} \quad (9)$

The characteristic impedance of a medium is given by

$$\eta = \sqrt{\frac{j\omega\mu}{\sigma + j\omega\epsilon}} = |\eta|\, e^{j\theta_\eta} \quad (10)$$

Magnitude of complex impedance, η can be given by,

$$|\eta| = \frac{\sqrt{|\mu/\epsilon|}}{\left\{1+\left|\left(\frac{\sigma}{\omega\epsilon}\right)\right|^2\right\}^{1/4}}$$

$$|\eta| = \frac{\sqrt{|\mu/(\epsilon' - j\epsilon'')|}}{\left\{1+\left|\left(\frac{\sigma}{\omega(\epsilon' - j\epsilon'')}\right)\right|^2\right\}^{1/4}}$$

$$\Rightarrow |\eta| = \frac{\sqrt{\mu_0/\epsilon_0}\sqrt{|1/(\epsilon_r' - j\epsilon_r'')|}}{\left\{1+\left|\left(\frac{\sigma/\omega\epsilon_0}{(\epsilon_r' - j\epsilon_r'')}\right)\right|^2\right\}^{1/4}}$$

$$= \frac{\dfrac{377}{\{(\epsilon_r')^2 + (\epsilon_r'')^2\}^{1/4}}}{\left\{1+\dfrac{(\sigma/\omega\epsilon_0)^2}{\{(\epsilon_r')^2 + (\epsilon_r'')^2\}}\right\}^{1/4}} \quad (11)$$

Phase of complex impedance can be given by,

$$\theta_\eta = \frac{1}{2}\tan^{-1}\left|\frac{\sigma}{\omega\epsilon}\right| \quad (12)$$

$$= \frac{1}{2}\tan^{-1}\frac{\sigma/\omega\epsilon_0}{\{(\epsilon_r')^2 + (\epsilon_r'')^2\}^{\frac{1}{2}}} \quad (13)$$

In this way the electromagnetic propagation through a medium is modeled.

IV. Power Calculations

Using the power relations given in [14-15] as a general rule for average power transfer per unit area through reflection and transmission about an interface separating two lossless dielectric regions 1 and 2, incident, reflected and transmitted power for every interface can be determined.

$$\text{Incident power,}\quad P_i^+ = \frac{(E_1^+)^2}{2\eta_1} \quad (14)$$

$$\text{Reflected power,}\quad P_r^- = \frac{(\Gamma E_1^+)^2}{2\eta_1} = |\Gamma|^2\frac{(E_1^+)^2}{2\eta_1} = |\Gamma|^2 P_i^+ \quad (15)$$

$$\text{Transmitted power,}\quad P_t^+ = (1 - |\Gamma|^2)P_i^+ \quad (16)$$

When the power carried by the radar wave is incident on any interface, 'n' separating the two tissue mediums 'n' and 'n+1', known as incident power during forward propagation, LP_{ni}^{+}, part of it is transmitted to the next layer in the same forward direction, known as transmitted power, LP_{nt}^{+} and the remaining power is reflected into its previous layer in the backward direction known as reflected power component, LP_{nr}^{-}. The amount of power reflected from every interface keeps getting retransmitted in a backward propagation mode and is finally received at the receiver. Such retransmitted power components from each of the interfaces received by the radar receiver in a backward propagation mode is known as reflected power component during backward propagation, $(LP_{nr}^{-})'$. The transmitted power or the reflected power respectively through or from every interface during any mode of propagation, forward or backward should be multiplied by the power attenuation factor of the corresponding layer before entering into the next tissue layer.

V. The Planar Models

Two planar techniques for modeling the propagation of electromagnetic waves (i.e. the UWB pulses) through human tissues are presented: the intrinsic impedance model (or classical model) and the impedance transformation model. It will be shown that both of the planar models can predict better signal attenuation as they take into account several important factors missing in the earlier models. However, the impedance transformation model predicts more subtle phenomena such as multiple reflections and hence is a more appropriate model. First, a five layer configuration: fat-muscle-cartilage-lung-heart is used to study attenuation of signal from heart and then the analysis is repeated for a six layer configuration: skin-fat-muscle-cartilage-lung-heart to know the attenuation contributed due to inclusion of skin. Both deflated and inflated lung types are used separately for a comparison.

a) Intrinsic Impedance Model

This is a multi-layered model having layers of different dielectric properties as depicted in Fig.4. The thickness and dielectric properties of the individual layers of the tissue structure are the constitutive parameters of the model. The velocity of propagation of a signal depends on the tissue permittivity and both permittivity and conductivity determine the attenuation of the signal. The reflection and transmission of signals take place at the interfaces between consecutive tissues and is obtained from the difference in impedance of the layers on either side of the interface. The intrinsic impedance of tissue layers separated by interfaces as shown in Fig.5 depends on the frequency dependent dielectric properties and hence is frequency dependent as well.

In general, considering a multi-layer system and assuming normal incidence, the E-field complex reflection coefficient of n-th layer is given by

$$\Gamma_n = \frac{E_{n_r}}{E_{n_i}} = |\Gamma_n|\, e^{j\phi_n} = \frac{\eta_{n+1} - \eta_n}{\eta_{n+1} + \eta_n} \tag{17}$$

where Γ_n is the reflection coefficient of the interface n between the layers n and $n+1$; η_n and η_{n+1} are the complex impedances of layers, n and n+1 respectively according to the concept of reflection of uniform plane waves at normal incidence. The electric field (E-field) reflection coefficients during forward propagation for every interface between consecutive layers of the model are obtained using the Eq.17 which will be useful for calculation of powers across the interfaces.

b) Impedance Transformation Model

In the event of wave reflection from multiple interfaces, impedance transformation method considers complicated sequence of multiple reflections in every region or layer as explained in [14]. Following the three-interface case related to the concept of wave reflection for multiple interfaces as shown in Fig.5 and using the boundary conditions at the tissue interfaces, we can have,

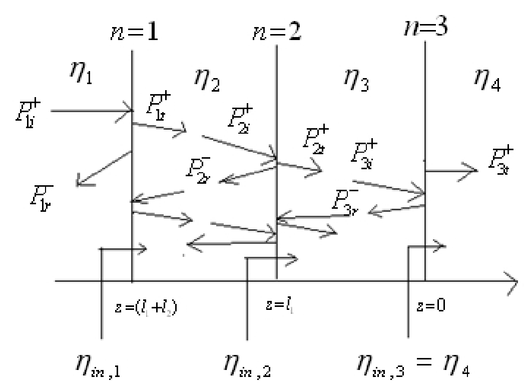


Figure 5 : Impedance Transformation Model

Input impedance at n=2,

$$\eta_{in,2} = \eta_3 \frac{\eta_4 \cos\beta_3 l_3 + j\eta_3 \sin\beta_3 l_3}{\eta_3 \cos\beta_3 l_3 + j\eta_4 \sin\beta_3 l_3} \tag{18}$$

Input impedance at n=1,

$$\eta_{in,1} = \eta_2 \frac{\eta_{in,2} \cos\beta_2 l_2 + j\eta_2 \sin\beta_2 l_2}{\eta_2 \cos\beta_2 l_2 + j\eta_{in,2} \sin\beta_2 l_2} \tag{19}$$

Now, reflection coefficient at interface n=1 can be given by,

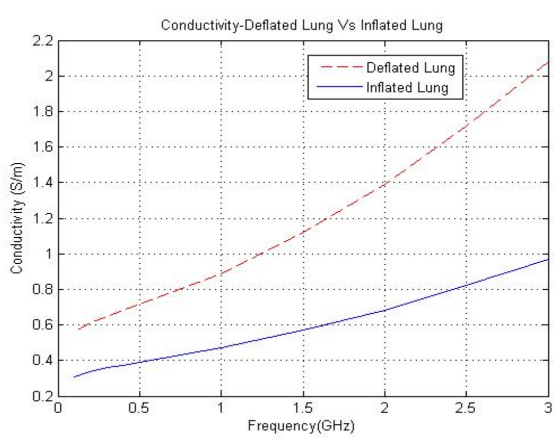
$$\Gamma_1 = \frac{\eta_{in,1} - \eta_1}{\eta_{in,1} + \eta_1} \tag{20}$$

Similarly, considering the six-layered tissue system of both of the models the input impedance and reflection coefficient corresponding to every other tissue interface are obtained using a MATLAB program and are then used for power calculations.

VI. Results

The backward reflected power from the heart i.e. the heart-lung interface of the air-heart-air path is the attenuation due to the UWB pulse echo for the frequencies in the band 0.1 to 3 GHz are calculated. This attenuation predicted by both classical model and impedance transformation model are obtained separately for the cases of deflated lung and inflated lung. Such attenuation results are also obtained for both lung types separately for without-skin and with-skin cases to know the contribution of skin tissue to attenuation. All attenuation results thus calculated are analyzed and compared with each other and also with the attenuation results reported by the circuital model [9] as well.

A comparison of conductivity, reflection coefficients at the lung-heart interface of both intrinsic impedance model and impedance transformation model is shown in Fig.6. It is clear that the conductivity of both deflated lung and inflated lung increase with frequency, the reflection coefficients of both lung types decrease with frequency. Also, at any frequency, deflated lung has higher conductivity and hence lesser reflection coefficient than inflated lung.



(a)

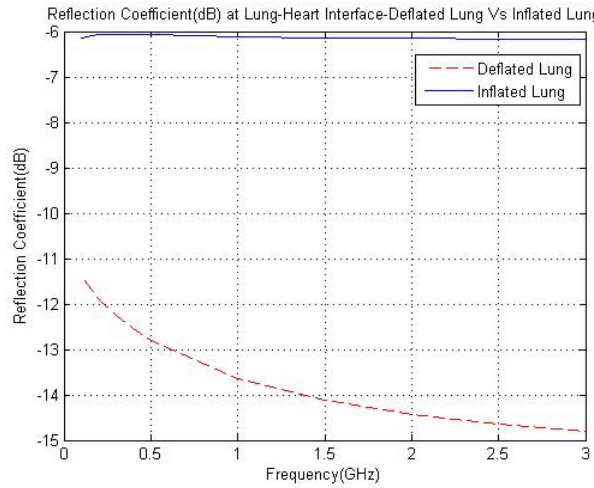
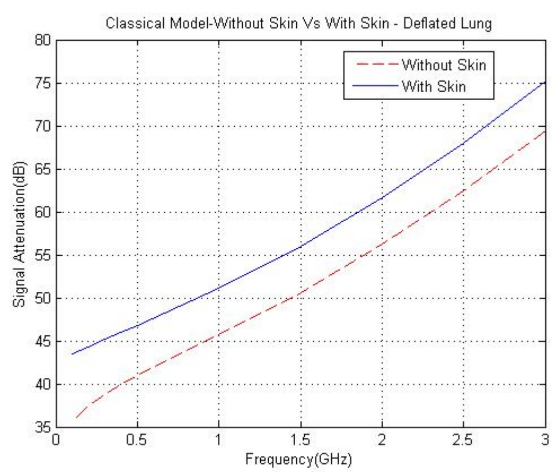


Figure 6 : Variation of (a) conductivity (b) reflection coefficient at the heart-lung interface – Deflated lung Vs Inflated lung

Attenuation increases with increase of frequency for both cases of without-skin and with-skin and moreover an additional attenuation in the order of 5 to 8 dB is found due to the inclusion of skin in the classical model [Fig. 7(a) and (b)]. Additional attenuation exclusively due to inclusion of skin is found in case of the impedance transformation model also [Fig. 8(a) and (b)]. However, in the later model, the amount of additional attenuation due to inclusion of skin is smaller at some frequencies as compared to that found in case of the classical model.



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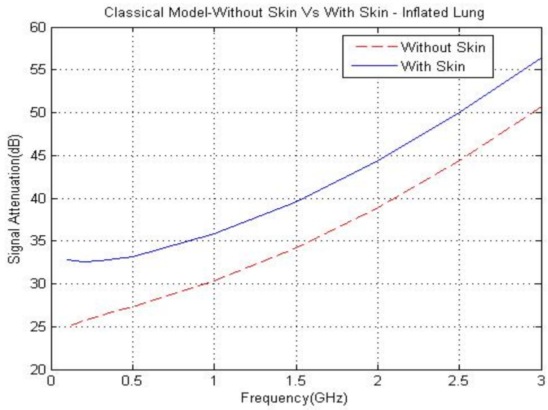
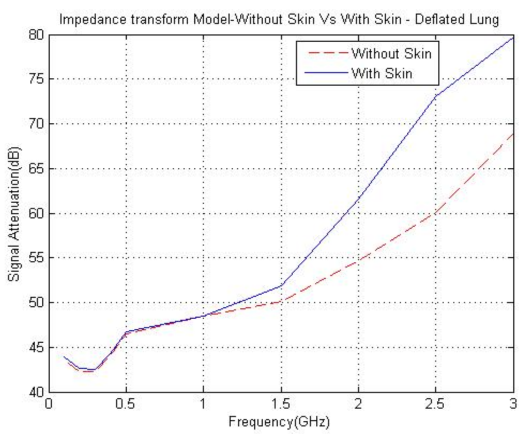


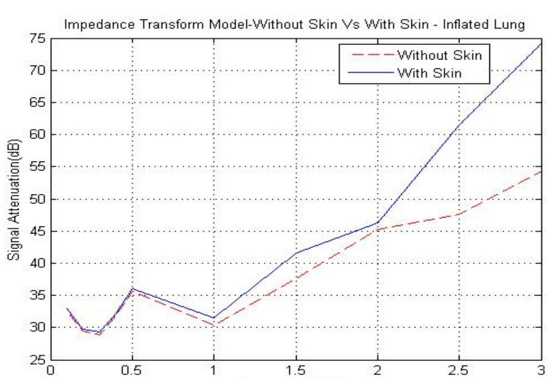
(b)

Figure 7 : Comparison of classical model predicted signal attenuation in decibels between without skin and with skin cases (a) deflated lung (b) inflated lung

The comparative analysis between deflated lung and inflated lung cases of both classical model as well as impedance transformation model each considering without-skin and with-skin cases are shown below in Fig. 9(a)-(d).

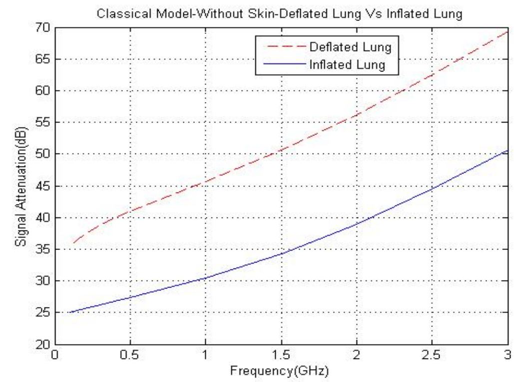


(a)

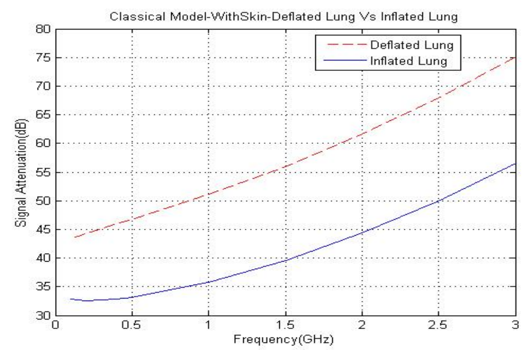


(b)

Figure 8 : Comparison of impedance transform model predicted signal attenuation in decibels between without skin and with skin cases (a) deflated lung (b) inflated lung



(a)



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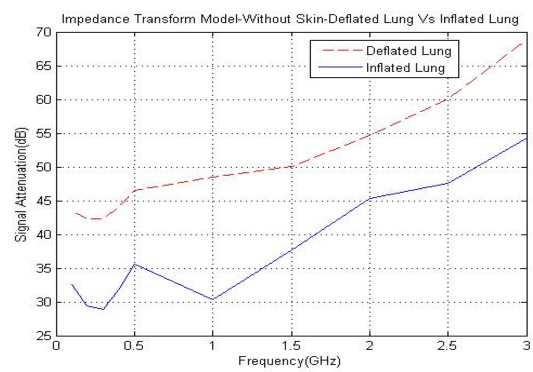
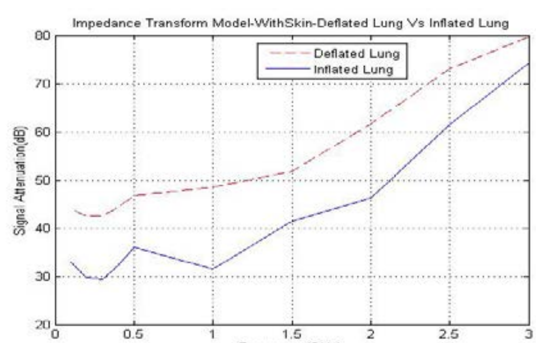


(c)



(d)

Figure 9 : Comparison of signal attenuation between deflated lung and inflated lung types for (a) classical model (without skin case) (b) classical model (with skin case) (c) impedance transform model (without skin case) (d) impedance transform model (with skin case)

In both of the models it is found that the increase of attenuation is higher in case of deflated lung than that in case of inflated lung for both without-skin and with-skin cases. This is due to the fact that deflated lung has higher conductivity [Fig.6(a)] and hence lesser reflection coefficient than inflated lung [Fig.6(b)].The amount of increase of attenuation from inflated to deflated lung at a particular frequency is the same between without-skin and its corresponding with-skin case of the same model. Such an increased attenuation characteristics from inflated to deflated lung was reported in the circuital model [9] but the models presented in this work predict more attenuation for both deflated and inflated lung cases.

VII. Conclusions

Remote monitoring applications based on detection of heart and breath rates are effective because it is difficult to suppress completely the heartbeat and respiration related subjects' motions. Also, the propagation of waves through the subjects' body is mainly governed by the electromagnetic characteristics of body tissues. The electromagnetic response of the human tissue is highly frequency dependent.

In this paper, both of the frequency dependent electromagnetic models of human body presented incorporate the electromagnetic properties of body tissues including the skin tissue corresponding to a 0.1 to 3 GHz ultra wide-band (UWB) radar. The lossy power coefficients at all tissue interfaces are evaluated during both forward and backward propagations of UWB pulses. The reflection from the cardiac structure i.e. the heart wall has been quantified and a comparative analysis on attenuation due to the pulse-echo intensity is presented. The complex reflection coefficients and complex impedances of tissue layers are considered in this one-dimensional analytical solution as they are important for the validity of power budget analysis. The advantage of the planar impedance transformation model is the implicit inclusion of the electromagnetic wave phenomenon such as multiple reflections, taking place during propagation through multiple interfaces.

Owing to the inclusion of the effect of skin tissue in the planar modeling along with the implicit provision for multiple reflections, the impedance transformation model can have better prediction ability of signal attenuation in the design of UWB radars. Future work will involve the application of impedance transformation model for other biological applications for study of signal attenuation. Results of such complex and sophisticated biological models shall help to build and test signal processing techniques in non-invasive, remote monitoring applications like home health care, emergency rooms, intensive care units (ICUs) in hospitals, pediatric clinics to alert for sudden infant death syndrome (SIDS), radar based lie detection, remote life detection etc. the heartbeat signal power thus derived can be processed further to obtain heart beat information.

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