Modelling the Interaction of THz Waves with Breast Cancer Tissues

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***Abstract*—We demonstrate the application of Double Debye theory combined with Finite-Difference Time-Domain technique to model terahertz wave interaction with breast tumor tissues. It has been shown that terahertz signals are strongly absorbed by water. Biological tissue has high water content, which is a possible reason for contrast when utilizing terahertz technology in breast cancer imaging. While results demonstrated the contrast between fatty and cancerous breast tissues, the challenge lies in the low contrast between cancerous and fibroglandular tissues (normal tissue). Therefore, computer simulations are needed to investigate a method to manipulate and enhance the contrast between these tissues for the sake of using terahertz technology in tumor margin assessment.**

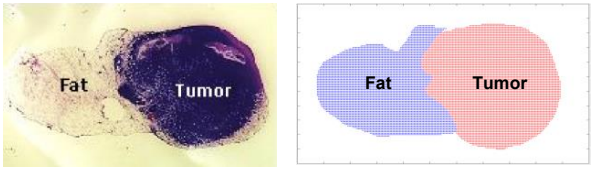
Keywords—terahertz (THz); breast cancer tumor; finite-difference time-domain (FDTD);

I. Introduction

As known, biological tissue is a dispersive medium in which the refractive index is frequency-dependent [1]. In this work, the breast tumor is modeled as a Debye dispersive heterogonous medium [2] that is incorporated into the finite difference time domain (FDTD) model. Double Debye parameters are calculated using the complex permittivity of breast tissues based on the measured experimental transmitted or reflected electric field. The computer simulations here use the Debye parameters to investigate the THz pulse propagation and reflection from the tumor tissue regions. Specifically, reflected data is collected and used for time domain THz image construction.

As known, the FDTD model requires spatial discretization at least 10 times smaller than the wavelength [1]. For 1 THz wave, the wavelength is 300 µm in air and smaller when the wave propagates inside biological tissues. This fact could lead to excessive need of CPU time for 3D configurations. Therefore, the approach based on the graphic processing units (GPUs) will be implemented here in order to speed up the computer code.

The configuration used here mimic the scanning mechanism of the THz system with an incident signal obtained from the experimental TPS 3000 system available at the University of Arkansas. We will use the computer simulations to visualize the reflected THz signals from fixed and freshly excised mice breast cancer tumors. The simulated THz images obtained using the far-field transformation will be compared with the experimental images obtained from scanning breast tumors, see [3], and also with the histopathology images (see Fig. 1). Understanding the way THz interacts with the tissue will contribute to advancing the experimental image construction.



(a) (b)

Fig. 1. Imaging of transgenic mice tumor (a) The histopathology image, and (b) digital configuration for the FDTD model based on the histopathology image in (a).

II. Methodology

A. Double Debye Theory

In the THz regime, studies have shown that breast tissues can be modeled by double Debye theory [2]. Debye model describes the frequency-dependent and complex dielectric function $\hat{\epsilon}(w)$ of a material by

$$\hat{\epsilon}(w) = \epsilon_\infty + \frac{\epsilon_s - \epsilon_2}{1+iw\tau_1} + \frac{\epsilon_2 - \epsilon_\infty}{1+iw\tau_2} \quad (1)$$

Where ϵ_∞ is the limiting dielectric constant at high frequency, ϵ_s is the static dielectric constant at low frequency, and ϵ_2 is the dielectric constant at the intermediate frequency limit. τ_1 and τ_2 are the slow and fast relaxation time constants, respectively [4]. In order to model the response of the breast tissue to THz radiation, we need to determine the Debye parameters. In this work, we use a curve-fitting approach to achieve Debye parameters to be used in the FDTD model. We use the measured complex refractive index $\hat{n}(w)$ of breast tissue at room temperature (298 K) between 0.1 and 2 THz from [2]. As

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explained in [2] we can calculate the complex dielectric constant of a material from refractive index and absorption coefficient by

$$\hat{\varepsilon}(w) = \varepsilon'(w) - i\varepsilon''(w) = \hat{n}(w)^2 \quad (2)$$

$$\hat{n}(w) = n(w) - i\frac{c}{2w}\alpha(w) \quad (3)$$

Where c is the velocity of light, $n(w)$ is the real refractive index and $\alpha(w)$ is the absorption coefficient of breast tissue. We use equations (2) and (3) to obtain the real and imaginary terms of $\hat{\varepsilon}(w)$. Therefore, we can fit the measured complex permittivity $\hat{\varepsilon}(w)$ to the permittivity of equation (1) for $\hat{\varepsilon}(w)$ and obtain five characterizing double Debye values. Here we use the least square fitting method to fit equations (2) and (3) into (1) for the given frequency range.

III. Results

A. Double Debye Values for the Breast Tissue

To test the Double Debye model, we use the measured refractive index and absorption coefficients of human breast tissues. The real part of the refractive index and the absorption coefficients, $n(w)$ and $\alpha(w)$, are shown in Fig. 2a [2]. The Deby model values of the permittivity are shown in Fig. 2b along with the calculated values using (2) and (3). As shown in Fig. 2b the Deby model values (fitted curves) are perfectly matched to the calculated permittivity of breast tissue. Table I shows the corresponding Debye parameters.

TABLE I. Debye Model Parameters

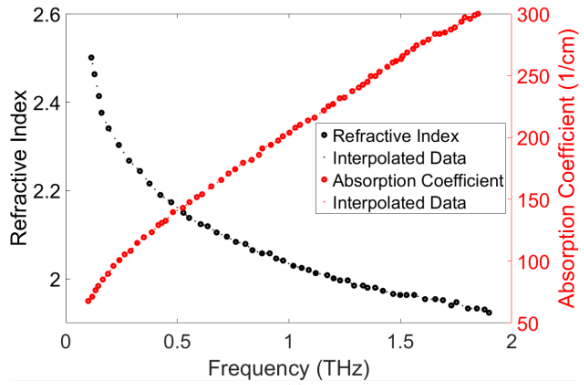
Debye Parameters	ϵ_∞	ϵ_s	ϵ_2	τ_1 (ps)	τ_2 (ps)
Values	2.98	5.88	5.43	2.08	0.25

B. Digitizing the Breast Tumor

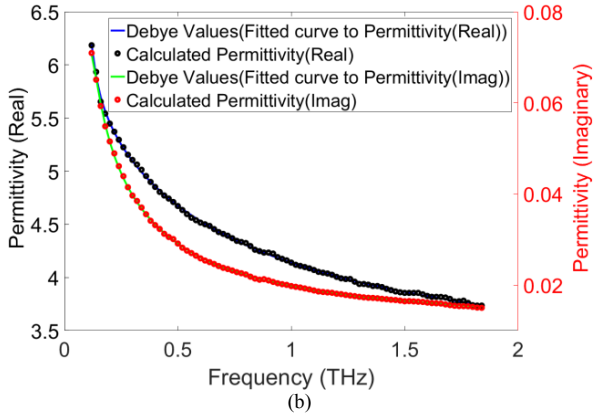
To be able to incorporate the breast tissue into the FDTD model, we digitized the breast tissue into different media according to Fig. 1a. We implemented an algorithm to differentiate and digitize the geometries contained in the tumor (see Fig. 1b). For more complex tumors that includes more than two regions, we can modify the algorithm and assign different Debye parameters to each region. In the mouse tumor in Fig. 1a, only two tissue regions are shown as Fat and cancer.

C. GPU Implementation

We will use NVIDIA Tesla K40 GPU to accelerate the FDTD model [5]. Tesla K40 is based on the NVIDIA Kepler GPU architecture. It contains 15 streaming multiprocessors (SMXs), each of which consists of 192 CUDA cores and 64 double-precision cores. Altogether, there are 2,880 CUDA cores and 960 double-precision cores, providing 4.29 Tflops single-precision performance and 1.43 Tflops double-precision performance, respectively. A Tesla K40 GPU contains 12 GB GDDR5 memory with a 288 Gbytes/second throughput.



(a)



(b)

Fig. 2. (a) Refractive Index and absorption coefficient of the breast tumor [2], and (b) Fitted curves of the permittivity of breast tissue.

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