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Millimeter and Submillimeter Wave Dielectric Properties of Tumorous and Non-tumorous Breast Tissues

Usman Khan[1], Nawaf Al-Moayed[1], Nicholas Nguyen[1], Mohammed Afsar[1], Stephen Naber[2]

[1]High Frequency Materials Measurement and Information Center
Tufts University, Medford, MA 02155, USA
[2]Dept. of Pathology, Tufts - New England Medical Center,
Boston, MA 02111

Abstract – **The refractive index, absorption coefficient, complex real and imaginary permittivity, and loss tangent of normal (non-tumorous) and tumorous breast tissues have been obtained from 0.2 to 0.9 THz. Dispersive Fourier transform spectroscopy was utilized to acquire the broadband millimeter and submillimeter wave dielectric properties and unique resonance signatures of breast tissues for the first time. Results indicate a significant difference in the electrical characteristics of tumorous and non-tumorous breast tissues.**

Keywords: **breast tissues, dielectric properties, dispersive Fourier transform spectroscopy, millimeter wave, submillimeter wave**

I. INTRODUCTION

For half a century, scientists have developed and improved several techniques for cancerous tissue detection and prognosis. 3-D Imaging methods such as magnetic resonance imaging (MRI), computed tomography (CT) scans, and mammography use electromagnetic radiation like RF waves and X-rays, to generate 3-D images of bones and tissues [1]. While these procedures are well-established, they are costly and repeated tests can ameliorate the risk of cancer [2]. Furthermore while imaging techniques can provide basic detection facilities, they do not yield information on the biological effects of electromagnetic exposure which is a serious concern given the explosion of sophisticated wireless and communication technology. To understand the effect of electromagnetic pollution on humans, one must analyze the dielectric characteristics of the human body. Dielectric properties are dependant on several parameters including the molecular composition and structure, surface area, frequency, and density; and thus serve as excellent tools for cancer detection.

Numerous techniques have been developed for the characterization of breast tissues. Early *in-vitro* studies of the electric properties of tissues and cells were conducted by Schwan [3-6]. Popovic, Okoniewski, Hagness, M. A. Stuchly, and S. S. Stuchly have published several papers on cancer detection using confocal systems, capacitive sensors, antennas, and probes [7-12] but their research is primarily limited to microwave dielectric spectroscopy which has several disadvantages. Probe and sensor measurements are extremely sensitive to surface level modulations and external environmental conditions. Unless ultra caution is exercised, inaccurate interface between the probe and sample can distort results. Microwave techniques are also roughly limited in the spectra with an upper limit of 30 GHz and lower limit of 300 MHz. This corresponds to wavelengths of 1 cm to 1 m respectively. While the penetration depth of such microwave signals is greater than millimeter and submillimeter wave signals, their resolution is also much less. Thus, while microwave signals may be able to carry more power and detect tumors further below the tissue surface, they can not obtain characteristics of tissues with greater accuracy and resolution compared to millimeter and submillimeter wave signals. Furthermore, using low power radiation at higher frequencies is desirable to avoid potential damage to human tissues.

To obtain a clear and thorough characterization of normal, benign, and malignant human tissues, it is important for one to use a technique that can yield highly accurate broadband dielectric data with the flexibility to accommodate different shapes, sizes, and composition of tissues. Dispersive Fourier Transform Spectroscopy (DFTS) provides the best alternative to broadband frequency domain examinations of human tissues. DFTS is a highly adaptable technique that allows variable temperature and variable sample size data acquisition from 60 GHz to 10 THz. This method has been used extensively for solid, liquid, and gaseous samples and can yield dielectric data with a sensitivity of 10^{-6} [13-18]. This paper shall discuss the utilization of DFTS for dielectric characterization of non-tumorous and tumorous breast tissues at the millimeter and submillimeter wave frequencies.

II. EXPERIMENTAL SETUP

A. Tissue Samples

The breast tissues were obtained courtesy of the Tufts-New England Medical Center. Tissues were removed from patients, fixed and preserved in 10% formalin containers, before being transported to the Dept. of Electrical Engineering where they were tested using a two-beam polarizing interferometer. The samples were cut to a size of approximately 2x1x0.5 cm^3. Sample holders were prepared to accommodate the tissues accordingly. The sample holders are 3"-diameter aluminum plates shaped like an annulus. Instead of having a circular cavity in the center, the opening is cut precisely to the dimensions of the breast tissue. The rectangular hole is then covered by a thin layer of mylar (thickness: 0.019 cm $\pm$ 0.001cm) on one side so that the tissue may rest on a flat surface.

B. Measurement System and Theory

DFTS was implemented using a two-beam polarizing interferometer. The setup for such a system has been discussed extensively [13-20]. In this system, a mercury-vapor lamp provides the source of radiation. The radiation is split by wire-grid beam-splitters sending part of the energy to the sample chamber and another part to a micrometer-backed moving mirror. After reflecting back from the scanning mirror and the fixed mirror in the sample chamber, the beams recombine and are collected by a liquid helium-cooled Indium Antimonide (InSb) detector. The reference run conducted includes the mylar-backed sample holder in the sample chamber. This yields a distorted reference interferogram, $V_1(x)$, which already takes into account the effect of the aluminum disk and mylar layer. A subsequent run with the tissue now in place is performed providing the interferogram, $V_2(x)$. The multiple reflections in the interferograms are non-existent due to the relatively lossy nature of tissues, thus both $V_1(x)$ and $V_2(x)$ can be expressed as a sum of their reflection and single transmission signatures as shown in (1). A double sided Fourier transform of the interferograms yields $F_1(\tilde{v})$ and $F_2(\tilde{v})$ respectively, the frequency domain data. Using Maxwell's equations and the dielectric definitions (2) and (3)

$$V(x) = V_R(x) + V_T(x) \qquad (1)$$

$$\hat{\varepsilon}(\tilde{v}) = \{\hat{n}(\tilde{v})\}^2 = \varepsilon'(\tilde{v}) - i\varepsilon''(\tilde{v}) \qquad (2)$$

$$\tan\delta = \frac{\varepsilon''(\tilde{v})}{\varepsilon'(\tilde{v})} \qquad (3)$$

one can obtain the refractive index (4) and absorption coefficient (5)

$$n(\tilde{v}) = 1 + \frac{x}{d_S} + \frac{ph\{\hat{F}_T(\tilde{v})\} - ph\{\hat{F}_1(\tilde{v})\} - ph\{(\hat{F}'(\tilde{v}))^2\}}{4\pi\tilde{v}d_S} \qquad (4)$$

$$\alpha(\tilde{v}) = \frac{1}{d_S}\left[\ln\frac{\hat{F}_1(\tilde{v})}{\hat{F}_T(\tilde{v})} + \ln(\hat{F}'(\tilde{v}))^2\right] \qquad (5)$$

where x is the shift, d_S is the sample thickness, $\tilde{v}$ is the frequency, and ph{} indicates the phase of the contents within the parentheses. $\hat{F}_T(\tilde{v})$ and $\hat{F}_1(\tilde{v})$ are the Fourier transforms of the sample and reference interference pattern respectively. $F'(\tilde{v})$ is derived from the ratio of $\hat{F}_T(\tilde{v})$ and $\hat{F}_1(\tilde{v})$.

III. Results

The refractive index, absorption coefficient, and complex real and imaginary permittivity, and loss tangent were acquired from 0.2 to 0.9 THz. Five tumorous tissues and five non-tumorous tissues were utilized to detect a general trend in the behavior of both sets of tissue. The results presented in Figs. 1 to 4 display the average properties of all ten tissues and indicate a clear difference between the two sets of tissue. The absorption coefficient is exhibited in Fig. 1. It can be seen clearly that the tissues with a tumor are much more absorbing than the non-tumorous tissues. A similar trend can be seen in the refractive index shown in Fig. 2, where the tumorous tissue has a refractive index value around 1.1 and the non-tumorous tissue has an average refractive index of 1.07 across the spectrum. Figs. 1 and 2 also contain several peaks. The first peak occurs around 311 GHz in both the tumorous and non-tumorous tissues. Around 460 GHz the tumorous tissue exhibits a broad peak where as the non-tumorous tissue is characterized by a very sharp peak. The 460 GHz peak can be attributed to the 10% formalin used to preserve and fix all the tissues. Previous studies have shown that 10% formalin has a broad characteristic peak in this region [13]. It is interesting though that while the tumorous tissue retains the broad peak, the non-tumorous tissue produces a much sharper peak. In addition to the difference of the average dielectric values between the two tissue types, the following two peaks can be used to differentiate a tumorous tissue from a non-tumorous one. A sharp peak characterizes non-tumorous tissues at 732 GHz as can be seen in Figs. 1, 3 and 4. Interestingly this peak dampens and shifts to 787 GHz for tumorous tissues. Figs. 3 and 4 display the real permittivity and loss tangent and confirm the dielectric behavior observed in Figs. 1 and 2. The real permittivity, ε', values for tumorous tissues are

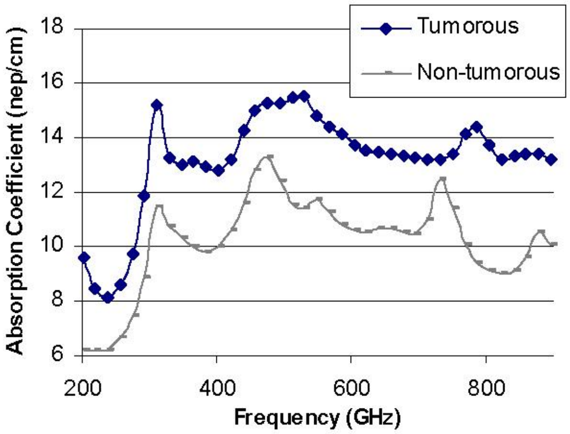

Fig 1: The absorption coefficient of tumorous and non-tumorous tissues from 200 to 900 GHz. Peaks can be observed at 311, 460, 732 and 787 GHz.

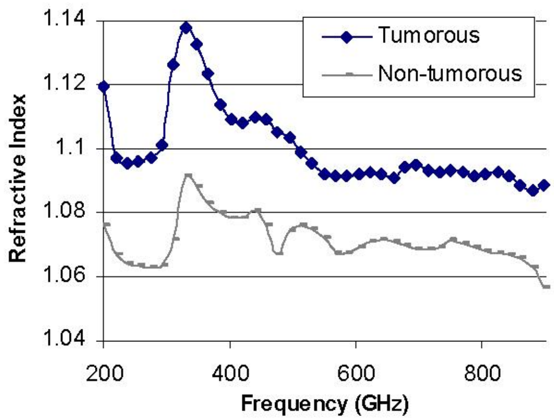

Fig 2: The refractive index of tumorous and non-tumorous tissues from 200 to 900 GHz. Peaks can be observed at 329 GHz and a dip at 476 GHz for non-malignant tissues.

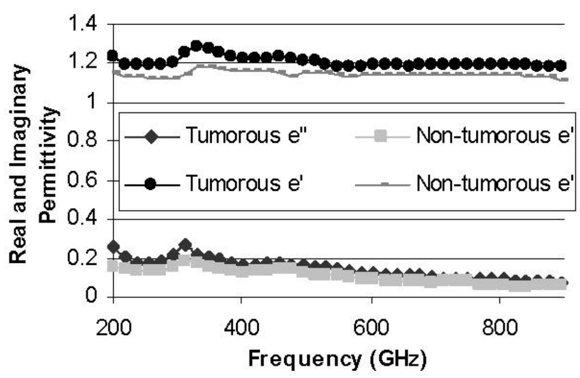

Fig 3: The complex real and imaginary permittivity of tumorous and non-tumorous breast tissues from 300 to 900 GHz.

approximately 1.2 where as non-tumorous tissues have a ε' around 1.15. The loss tangent indicates that at higher frequencies the loss decreases.

IV. Discussion

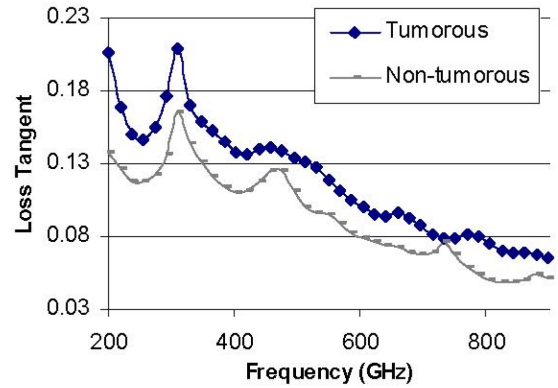

Fig 4: The loss tangent of tumorous and non-tumorous tissues from 200 to 900 GHz. Peaks can be observed at 311, 460, 732 and 787 GHz.

Breast tissues have been studied previously at lower frequencies using several techniques [7-11]. This research confirms the trend observed in these studies that tumorous tissues have significantly higher permittivity than non-tumorous tissues. To the best of the authors' knowledge, this is also the first time broadband millimeter and submillimeter wave dielectric properties of breast tissues were acquired and the unique resonance signatures identified. The 311 GHz peak in the absorption spectrum was present in all the breast tissues examined. After analyzing other tissues such as the lung, liver, or brain, its can be determined whether this is a signature unique to only the breast or common in other human tissues. The peak at 732 GHz was observed in all the non-tumorous tissues at different intensities. However the peak at 787 GHz for tumorous samples as seen in Fig. 1 did not appear in all the samples indicating the possibility of a localized resonance process in certain tissues. This could be attributed to the presence of air gaps or cavities in certain tissues.

Breast tissues are composed mainly of fatty and fibroglandular tissues. Many known cancer detection techniques such as mammography have limited capacity to distinguish these regions from one another for tissues with less than 45% fat [21]. DFTS operates in a region with very small wavelengths and can offer much greater resolution and accuracy for identifying the tumorous and non-tumorous, regions or fatty and fibroglandular tissues. The first step towards isolating these regions through dielectric mapping of the tissue has been in achieved with this research. With the average dielectric characteristics of tumorous and non-tumorous tissues known, one can investigate smaller tissues with multiple regions (malignant, benign, or normal tissue).

Typically DFTS measurements of solid materials have less than 1% random error [16]. However, because of the non-homogeneous nature of human tissues, uneven specimen surface, and 10% formalin's tendency to oxidize and evaporate, the random error in these breast tissue measurements is a maximum of 3.6%. The error can be significantly reduced if the breast sample size is increased and a larger number of samples are available for averaging purposes. A larger tissue diameter will enhance the measurement of permittivity and loss tangent at longer wavelengths. Furthermore, if greater precaution is exercised, one can decrease surface reflection losses by providing samples with a level surface. The evaporation rate for 10% formalin can also be reduced if the system temperature is lowered. A systematic error analysis can not be conducted since no published data exists for the dielectric properties of breast tissues in this spectral range.

V. Conclusion

The absorption coefficient, refractive index, complex real and imaginary permittivity of tumorous and non-tumorous tissues have been presented from 0.2 to 0.9 THz. While the behavior of both tissue types is similar, tissues with a tumor have greater loss, absorption, and refractive indices. Frequency signatures for breast tissues were detected and can be used in future to differentiate normal, benign, and malignant tissues.

In this study, several parameters such as the tissue composition, water content, and tumor size were not controlled. Subsequent studies are planned to understand the contribution of these parameters to the dielectric properties at the millimeter and submillimeter wavelengths. In addition to controlling these parameters, the authors hope to conduct an extensive study with a significantly larger number of samples from select demographic groups to identify the changes in the dielectric properties as a function of a women's age. Furthermore, it is hoped that other human organs such as the liver, brain, and breast tissues will be studied to analyze and estimate the effects of electromagnetic radiation on the human body.

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