COMPUTED THERMAL IMAGING FOR TERAHERTZ RADIATION MAPPING ON TISSUE SURFACE TEXTURE

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Accepted: 1 August 2019 | Published: 15 August 2019

Abstract:
THz power density-based thermal imaging has been explored to the image of biological tissue theoretically. To rely on mapping the relative temperature and thermal parameters noninvasively, we propose the THz power density-based thermal thermography in application of imaging for a texture of tissue surface. Two-dimensional thermal tomography images of biological tissue were obtained with using a power density range of (10-250) mW.mm⁻³ and a frequency range of (0.1-1) THz. The axial and lateral resolutions are characterized. Theoretically computed thermal tomography on the transient 1D heat conduction uses several types of biological tissue in the millisecond to picoseconds time range. The results for time intervals of one second or longer show a constant temperature or a steady state centered about one temperature. By contrast, millisecond to picoseconds time ranges display a small but significant temperature change as the depth varies about 0 °C which correlated with the contrasting tissue structures. The steady-state body temperature for second range and the transient-state for significant small change in milliseconds-picoseconds range toward thermal equilibrium is not restricted to be axial and radial depth or spatially invariant. THz power density-based thermal tomography irradiates surface texture of muscle-tumor tissue that gives a good quantitative description of geometrical structures. Morphology of tissue provides boundaries of images, their skeletons, and many preprocessing and post processing techniques, especially in edge thinning and pruning. The primary application of morphology occurs in binary images. The vertices of the tissue surface texture graph are highlighted and its adjacency matrix and several parameters of the graph are also displayed. This characteristic is physiologically realistic and technically accessible.

Keywords: thermal imaging, THz radiation, mapping, tissue surface

1. Introduction

The problems arising from the behavior of the relaxation time appears very clearly in the conduction equation through biological tissue at short time scales. The response to a localized THz radiation has shown that the temperature field is distributed instantaneously at every point of space from the moment the pulse comes into effect (Rui Chen DDS, 2018). It basically amounts to saying that computed thermography has neglected the response time of the system. Of course, the response to a temperature gradient imposed suddenly in this way cannot be instantaneous. In fact it is obvious that time scale of the response to a step shaped excitation; it is the time required for equilibrium to be reestablished, i.e., the relaxation time (Choun Pei Wong, 2018). THz radiation electric field generates an oscillating current and the rapid transfer of the energy of this current into the molecular motion responsible for most of the heat capacity results in an increase in the local temperature. Consequently, in short time
we temp to employ the computed thermal thermography in THz imaging computation on the tissue surface of a 2D texture to predict the temperature distribution and to map the heat conduction as the visualization in causal interpretation of energy density flow absorbing through two tissue layers. The results for time intervals of one second or longer show a constant temperature or a steady state centered about one temperature. By contrast, millisecond to picosecond time ranges display a small but significant temperature change as the depth varies which correlated with the contrasting tissue structures.

This work uses THz power source in 10-150 mW and samples of cow fat and muscle tissue while the theoretical work frame uses skin and tumor tissue. We directly map the computational imaging to dots, in some tissue surface areas the data could be sparse, while in other areas the plots overlap, leading to a flat appearance and loss of detail. This technique can apply Gaussian blur, and colorizing to achieve visually enhanced 2D rendering of certain datasets. A variation of the mapping is visually effective rendering a common challenge for visualizing fractals and other datasets. In computed thermography THz imaging the data density varies very significantly. It suggests one possible and simple approach to enhance the visual appearance to some extent, using density for thermal mapping, resolution adjustment, Gaussian blur, and colorization. Use of tissue image histogram to image enhancement operations shows significantly different sharp peak spectrum between normal and abnormal tissue surface. They have displayed the different thing about cancer and normal tissue image contrast. The purpose of heat mapping is to visualize the relationship between heat transfer rate and THz power density that varies thermal conductivity of tissue kinds by applying low scattering factor. This visualization plots the heat production rate versus wavelength through the depth of tissue. This technique is useful for calculating the two-dimensional electric fields of THz radiation source: the curve in the w-plane where either part of real or imaginary is constant corresponds to either an equipotential line or radiation electric flux (Ja-Yu Lu, 2005). The thickness of the shape is proportional to the energy density and the energy flux intensity is mapped with colors on this shape. In the construction of computed thermograph technique for a 2D texture of tissue images from slices on surface, we apply mapping of conformal technique and imaging computation on tissue images to use texture coordinates align with the x and y coordinates for non-parametric surfaces and regions. Parts of tissue surface indicate on components of texture are such as the morphology, vertices and adjacent matrix. To use a transform in the complex function of a complex variable, so that the analytical function maps the complex z plane into the complex w plane.

Image histograms are important techniques for application in THz imaging processing, most notably compression, segmentation, and thresholding. The part of histogram column it shows the effect of absorption spectrum color curve or peak shift adjustments on a computed thermography of image and its histogram.

2. The Surface and Boundary of the Abnormal – Normal Tissue Layer

THz radiation regime fields can be quantitatively described in terms of a “nonlinear” analysis. That is, it is assumed that the change in the density of the cancer cell tissue layer is linearly proportional to the change in the pressure which waves are of infinitesimal radiation field intensity amplitude. Radiation pressure a can exert steady forces on interfaces between tissue layers having different values of radiation velocity and/or density. In Figure 1a it shows a radiation penetration with scattering factor. Figure 1b and c, use of Maxwell equation solves the first boundary conditions at the interface between two tissue layers having forms of spherical and cylindrical geometry,
respectively. It uses the important integral theorem or a divergence theorem. If this theorem used on the surface integral of magnetic field normal line component, then each of part of this integral is zero. Detail explanation each boundary of space is given

\[ \int_{S_1} \hat{n} \cdot B \, da + \int_{S_2} \hat{n} \cdot B \, da + \int_{S_3} \hat{n} \cdot B \, da = 0 \]  

(1.1)

Now in Fig.1.1c it lets \( h \to 0 \)

\[ \int_{S_1} \hat{n} \cdot B \, da - \int_{S_2} \hat{n} \cdot B \, da = 0 \]  

(1.2)

n’s are in opposite directions. At interface is obtained the requirement: \( B_{1n} = B_{2n} \).

Analysis on the surface part results in the equation

\[ \hat{n} \cdot \nabla \times E \, da = -\frac{\partial B}{\partial t} \, da = 0 \]  

(1.3)

Eq. (1.3) relation to electric field intensity amplitude \( E \) in the closed line integral with the surface integral of magnetic field \( B \) in normal direction. Tangential component of electric field intensity amplitude \( E_t \) is continuous in interface. The left part of Eq (1.3) may be resulted in the next equation expression

\[ \frac{\partial E_{1t}}{\partial t} - \frac{\partial E_{2t}}{\partial t} + h_1 E_{1n} + h_2 E_{2n} - h_1 E_{1n} - h_2 E_{2n} = -\frac{\partial B}{\partial t} \, da \]  

(1.4)

Now shrink loop, letting \( h_1 \) and \( h_2 \to 0 \). So \( h \) terms vanish and so does right hand side, provided only that \( \frac{\partial B}{\partial t} \) is bounded so \( \frac{\partial E_{1t}}{\partial t} - \frac{\partial E_{2t}}{\partial t} = 0 \). For \( \nabla \cdot D = \rho \) and letting \( h \to 0 \), \( (D \hat{n})_A + (D \cdot n) A = \sigma A \) is obtained \( D_{1n} - D_{2n} = \sigma \), now letting \( h \to 0 \), they are obtained some equations such as \( J_{1n} \cdot J_{2n} = -\frac{\partial n}{\partial t} \), \( D_{1n} \cdot D_{2n} = \sigma \) and \( J_{1n} \cdot J_{2n} = j\omega \sigma \). If light is monochromatic and \( \sigma \) varies as \( e^{-j\omega t} \), then Eq (1.4) can be expressed in electric field intensity \( E \) on interface of two tissue layer which permittivity \( \varepsilon \) connecting with dielectric constant \( K \) in \( \varepsilon = Ke \). The magnetic field equation in terms \( \to 0 \) when loop shrink for parts from \( H_{1n} \) to \( H_{2n} \) and rest of from the previous reason \( B_2 = 0 \), \( H_{21} = 0 \) and dielectric constant does not depend on frequency, then ratio
\[
\frac{H_{1t}}{H_{2t}} = \frac{h}{2} \left( \sigma_1 - j\omega K_{10} \right) + \frac{h}{2} \left( \sigma_2 - j\omega K_{20} \right) \frac{E_{20}}{E_{1n}}
\]

(1.5)

It calls \( h/2 \) the skin depth. So \(( h/2 ) \sigma_2 E_{2n}\) is the total current \( J_{S_{\perp}} \) Maxwell equation which is separated by the real and imaginary parts of a complex solution. However, equation of energy density and flux per area unit are nonlinear in radiation field. Energy flux \( S \) in interface of two biological tissue layers has been derived with using some approximations. Energy flux in W/m² sec and at the higher frequencies such as THz radiation range penetrating in biological tissue layers for its depth and heat transformation depends on kinds of tissue layers. If THz radiation electric field intensity amplitude has a source from an oscillating electric dipole, located at the origin then the radiated power unit solid angle \( S \cdot \hat{r} \cdot r^2 \).

3. Experiment Set Up And Modeling In Thz Imaging

In principle there are two types of THz radiation engineering: pulsed and continuous-wave. A pulsed system is established on the application of electromagnetic pulses in picoseconds or sub-picoseconds duration. This pulse impinges on a sample and the transmitted or reflected waveform is coherently recorded in the time-domain in Figure 2a. In transmission-mode pulse imaging set-up, as shown in figure the beam from the femtosecond laser is split before arriving at the emitter. One portion is incident on the emitter and generates Terahertz; while the other portion arrives at the dipole detector after passing through a delay line and being reflected from a number of mirrors, and helps through the procedure of Terahertz detection. The reflected THz beam is then focused on the object. After passing through the sample, it is again collimated. At this stage, after the interactions the photocurrent of the detector dipole is detected through a current amplifier and is shown on a lock-in amplifier which is connected to a PC in case of any need for data transfer or data recording (Irene Burghardt ,2009). Note that the data captured in this case is the photocurrent; therefore the power of the Terahertz would be proportional to the squared valued of this data. In this set-up the source and detector are at fixed positions but since the Terahertz emitter is a point source and the object must be imaged all over its surface, a motorized stage has been developed in order to raster scan the object.
Therefore, the only moving parts of this set-up are the object and the delay-line. The process stages of the THz pulse imaging set-up is shown in figure 2b. In the THz imaging described the sample is placed in the focal plane of the THz beam, however due to the geometry of the optics of the system the divergence of the beam is negligible. Therefore the approximation of normal irradiance is valid. The multilayered tissue analysis is applied to a series of slabs of tissue with homogeneous physical and optical properties in Figure 2c. It uses Maxwell’s equations for the THz radiation passing through tissue. The mathematical formulation presented assumes normal irradiance of the incident THz radiation on the sample. The dimensions of the layers and optical properties of the sample will be predetermined and any number of layers is feasible within the theoretical framework, as a scattering factor is calculated from parameter for each layer of the sample. Hence the layers of tissue can be built up assuming that the interface between them is linear. It is obviously no facility within the model to complicate the geometry of this interface, however it may not be important as the deviation of linearity in tissue is much smaller than the incident THz radiation wavelength. The system used in the analysis comprised three layers of medium through which the THz pulse transverses. Both the transmitted and reflected THz waveforms are modeled using the expressions given in multiple reflections in tissue. This analysis is performed for each frequency in the THz spectrum obtained by the Fourier transformation (Tom A. Waigh. 2007).

4. Results and Discussions

4.1 Thermal Mapping through Biological Tissue

In a visualization of causal interpretation for energy density plot it is observed through trace of THz radiation photon in tissue using Monte Carlo method. In Figure 3a the dark shading color agrees room temperature about 25 °C. Consider heat transfer given by heat conduction equation which represents heat conduction mapping in a two-dimensional domain. The two-dimensional system can be visualized directly with area rendering using a two-dimensional colormap and mapping in the amplitude to the opacity. The area in x
and y coordinate is in which at an origin point as a beginning position of radiation source trace that travels randomly in tissue to form a trajectory going from bottom to top, the radiation axial position x from left to right, and the radial distance r from bottom to top. One clearly sees the small random oscillations of the radiation trace toward top in radial direction. Such kind of visualization is not easy to interpret, particularly if it is statically depicted; with interactive slicing, however, one can analyze the structure of the wave function. Another way to visualize this data is to look at observables such as the energy density as well as the flux density and to transform the system back from its mathematical space into real space. This gives type of quantities which are visualized simultaneously in energy density and the energy flux intensity mapped with colors on shape.

In Fig. 3b, 3c and 3d the boundary conditions are such that the temperature on all the edges of the domain equal to 0. Without loss of generality one can take the thermal diffusivity of skin, fat and muscle is α equal to 0.05 cm$^2$/s, 0.007 cm$^2$/s and 0.003 cm$^2$/s, respectively.

The initial condition is given in room temperature. The dimensionless temperature can be found using the Chebyshev collocation technique Choun Pei Wong (2018). They give the results in different contours of the dimensionless temperature and show using the various colors for high temperature in red and low temperature in orange. Perfect agreement is observed which fat indicates higher in heat transfer than skin and muscle. It refers on difference of tissue characteristics. This analyzes uses various tissues with a power source of 100 mW. Fat shows high heat production rate (red) among skin and muscle.

4.2 Imaging Computation of THz Radiation Absorption through Tissues

In Table 1 the part of columns illustrate comprehensively components of stage in the construction of computed thermograph or tomography technique for a 2D texture of tissue images from slices on surface. Texture coordinates align with the x and y coordinates for non-parametric surfaces and regions. Each of column part it indicates the result of computation in imaging from top to bottom column: cancer skin, normal skin, normal muscle and tumor muscle, respectively. We use a transform in the complex function of a complex variable, so that the analytical function maps the complex z plane into the complex w plane. This technique is useful for calculating the two-dimensional electric fields of THz radiation source: the curve in the w plane where either part of real or imaginary is constant corresponds to either an equipotential line or radiation electric flux.
This technique can apply Gaussian blur, and colorizing to achieve visually enhanced 2D rendering of certain datasets (Choun Pei Wong, 2018).

A variation of the mapping is visually effective rendering is a common challenge for visualizing fractals of tissue and other datasets. In many cases the data density varies very significantly. If we directly map the computational solution to dots, in some areas the data could be sparse, while in other areas the plots overlap, leading to a flat appearance and loss of detail. In Table 2 tissue image histogram is an important concept common to many image enhancement operations is that of a histogram, which is simply a count (or relative frequency, if normalized) of the gray levels in the image. Analysis of the histogram gives useful information about cancer and normal tissue image contrast. Image histograms are important techniques for application in THz imaging processing, most notably compression, segmentation, and thresholding. The part of histogram column it shows the effect of absorption spectrum color curve or peak shift adjustments on a computed tomography of image and its histogram. The color curve shift defines the brightness relationship between the input (original) values (on the x axis) and output (result) pixel values (on the y axis). The histogram displays the absorption spectrum distribution of the normal and cancer tissue image. It displays level of brightness on the x axis from dark to light in table on columns tomography, histogram and Fourier we produce them using of the imaging computation. This technique applies the computed tomography with using the THz frequency range, Fast Fourier transform (FFT) and the inverse transform on the selected tissue images with a number of projection angles. In table they are the original tissues are given such as skin, fat, muscle and tumor. They show that the reconstructed tissues image start to resemble the original tissues as the number of projection angles increases. In principle of THz imaging the interaction of THz radiation with tissue can be correlated to the discontinuities of optical properties at the interface between two different tissues and the related volume on the optical properties of the tissues. These indications are bases for any. imaging technique. The tissue image targets can be identified by selecting narrow frequency band for 0.5 THz and .1 THz as a cut off frequency around absorption peaks of each tissue (see magnitude spectrum in the fourth column).

Fat and skin tissue image result in a spectrum of the sharp peak while muscle and tumor show spectrum of a various small peak. It is caused by using Fourier transform or the inverse transform to simulate this method in the maximum number of 128 projections can be taken between angle of -90° and 90°. Also applying the inverse Fourier transform on the resulting image gives the filtered back-projection image. Principle in transmission mode of THz imaging computation, the THz signal transmitted through the tissue is computed on the other side by a receiver where all etalon effects (multiple reflections) are ignored. A typical THz imaging Fourier transform computed each tissue type such as skin, fat, muscle and tumor tissue in area of row (x) row(y) is shown in Table 2.
Table 1 Construct a 2D texture on surface from slices of tissue images for non-parametric surfaces and regions

<table>
<thead>
<tr>
<th>Original Tissue</th>
<th>Texture of Surface</th>
<th>Component of Texture</th>
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<td></td>
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<td></td>
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<td></td>
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<td></td>
<td></td>
<td>1 component of size (328)</td>
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<td></td>
<td>vertices = 556</td>
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<td></td>
<td></td>
<td>edges = 796</td>
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<td></td>
<td></td>
<td>1 component of size (556)</td>
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<td></td>
<td>vertices = 463</td>
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<tr>
<td></td>
<td></td>
<td>edges = 651</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1 component of size (463)</td>
</tr>
</tbody>
</table>

The THz imaging magnitude spectrum of the incident THz signal is computed using the spectral resolution after the discrete Fourier transform is inversely proportional to the width of the incident signal, which is decided ultimately by a time window. Magnitude spectrum of skin and fat are sharp peaks while muscle and tumor are rather flat. It means that in factor of reflection mode, the situation is more difficult, as there are no longer sharp absorption peaks. Also the reflected electric field at normal incidence from a thick
sample is determined by the incident signal spectrum. All multiple reflections have been ignored. The reflection coefficient from the tissue is given by absorption spectrum is highly dependent on the imaginary portion, while the reflection spectrum depends primarily on the real portion which is simply the square of the index of refraction. The larger the number of projections applied on the original tissue image, the more accurate the reconstructed tissue image becomes. In computed tissue tomography, many projections of the tissue object are first generated from different angles. Then filtered back-projections are applied to reconstruct a 2D image of the structure of a particular cross section of the tissue image. This is the basic idea used in THz–radiation medical imaging.

Table 2 Computed Thermograph of a 2D texture on surface from slices of tissue images for non-parametric surfaces and regions

<table>
<thead>
<tr>
<th>Original Tissue</th>
<th>Thermograph (50 mw)</th>
<th>Histogram</th>
<th>Fourier Inverse Transform (Image Magnitude Spectrum)</th>
</tr>
</thead>
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</tr>
</tbody>
</table>

5. Conclusion

In the computed thermography with THz imaging the results for time intervals of one second or longer show a constant temperature or a steady state centered about one temperature. By contrast, millisecond to picoseconds time ranges display a small but significant temperature change as the depth varies for each tissue such as skin, fat, tumor, or muscle. It is correlated with the contrasting tissue structures. For computed thermography of THz imaging applies imaging computation with using Fourier transform and discrete Fourier transform with low
filter in frequency domain of the image which gives the result of blurring at areas of rapidly changing pixel intensity in tissue which can be correlated to the discontinuities of optical properties at the interface between two different tissues and the related volume on the optical properties of the tissues. The tissue image targets can be identified by selecting narrow frequency band for 0.5 THz and .1 THz as a cut off frequency around absorption peaks of each tissue. Fat and skin tissue image result in a spectrum of the sharp peak while muscle and tumor show spectrum of a various small peak. It is caused by using Fourier transform or the inverse transform to simulate this method in the maximum number of 128 projections can be taken between angle of -90° and 90°. Also applying the inverse Fourier transform on the resulting image gives the filtered back-projection image. Principle in transmission mode of THz imaging computation, the THz signal transmitted through the tissue is computed on the other side by a receiver where all etalon effects (multiple reflections) are ignored.

Acknowledgments

The authors thank the Department of Physics, Faculty of Math and Science, University of Riau Pekanbaru Indonesia and Department of Physics, Faculty of Science, Universiti Teknologi Malaysia, Johor Baharu for supporting in lab. facilities of this project.

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