초단파 펄스 트레인 레이저 조사시 생체 조직의 열반응

The Thermal Response of Biological Tissue Subjected to Short-Pulsed Irradiation Train

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1. Introduction

The most easily quantified and commonly observed mechanism is the photo thermal effect in laser tissue interaction [1]. The application of photo thermal effect is laser surgery [1], laser peri-implant treatment [2], laser induced interstitial thermotherapy [3], and so on. The focused laser beam is often used to ablate tumor tissue selectively [4] by irradiating tumor. In this paper, the hyperbolic conduction equation is introduced for the bio-heat transfer model. The radiation energy absorption is investigated and the temperature increment is expected. Most of previous result, the temperature prediction is modeled with parabolic conduction model. However, in the recent papers [5], the hyperbolic heat transfer in the biological medium is highlighted. Moreover, the laser radiation transport is solved with the TDOM (Transient Discrete Ordinate Method) in this paper. The non uniform grid system is then employed to capture the focused laser beam movement.

2. Mathematical models

2.1 Governing Equation

The pulses of a focused laser beam are incident tissue into 2 mm below the tissue surface as shown in Fig.1. Three layered tissue is modeled with inhomogeneity medium, which is modeled with a high scattering medium. The tissue optical and properties are summarized in the table 1.

The heat transfer model is formulated as a two dimensional axisymmetric model as:

$$\rho C_p \frac{\partial T(r,z,t)}{\partial t} = -\nabla \cdot \left[\mathrm{q}_{cond}(r,z,t) + \mathrm{q}_{rad}(r,z,t) \right] \tag{1}$$

where ρ is the density, C_p is the heat capacity, T is the temperature, $\mathbf{q}_{cond}$ is the conductive heat flux vector, and $\mathbf{q}_{rad}$ is the divergence of radiative heat flux vector. In hyperbolic thermal wave theory, the conductive heat flux vector is expressed by [5]:

$$\tau \frac{\partial \mathbf{q}_{cond}(r,z,t)}{\partial t} + \mathbf{q}_{cond}(r,z,t) = -k\nabla T(r,z,t) \tag{2}$$

where the thermal relaxation time τ and thermal conductivity k are introduced. Introducing the thermal diffusivity $\alpha = k/\rho C_p$, the speed of thermal wave is

$$c_t = \sqrt{\alpha / \tau} \tag{3}$$

Eq. (2) regresses to the traditional Fourier expression when $\tau \to 0$, and $c_t \to \infty$. To achieve the radiative heat flux vector, transient radiative transfer equation is solved with a 10 ps irradiation. Detail equations and solution technique about it were introduced previous paper works [5].

2.2 Numerical schemes

The Transient Discrete Ordinate Method (TDOM) is adopted to solve the transient radiative transfer equation and the MacCormack's predictor-corrector scheme is employed to solve the hyperbolic conduction equation. In current paper work, the influence of grid systems is highlighted. For the uniform grid system, both radial and axial distances are evenly divided. Two kinds of nonuniform grid systems are implemented as:

$$R_{i+1} = R_i + \Delta R \tag{4a}$$

$$\Delta R_i = \alpha_r \left(\beta_r - \exp(-\frac{\gamma_r i}{N}) \right) \tag{4b}$$

Z directional grid is implemented as a similar manner of radial direction. The grid parameters selected as below:

(Nonuniform gridI) α_r=0.0328; β_r=2;γ_r=10;N=201

(Nonuniform gridII)α_r=0.0427;β_r=1.2;γ_r=25; N=201

The fine grid is employed in the laser beam deposition area and coarse grid is employed to other region. The total number of grid is same as the uniform grid system, which means the calculation cost is not sacrificed.

3. Results and discussion

In Figure 2, the contours of divergence of radiative heat flux are depicted at selected time instants. At early time instant (t = 20 ps), the radiation energy absorption is mainly confined near the focal region. As time advancing, the field of radiation energy absorption propagates and is enlarged almost whole tissue medium. However, the maximum intensity of radiation energy absorption drops to about 10% at t = 200 ps. At the tissue surface region (Epidermis), high radiation energy absorption is predicted due to the strong light absorption characteristics.

In Figure 3, the temperature prediction is compared with experimental data along the z-axis after 10 sec. Generally, the data between numerical and experimental one is well matched. The predication result with nonuniform grid systems show the closer value with the experimental one rather than uniform grid system, especially peak temperature location. It seems no significant difference between nonuniform grid systems. The prediction of maximum temperature increment is most important in the laser surgery because it is the key parameter to estimate of ablation efficiency. The thermal wave damping motion is predicted in the simulation model. Initially, the high temperature gradient takes place between epidermis and dermis tissue region. The epidermis has high light absorbing characteristics, which causes the strong temperature concentration field. As time advancing, the thermal wave moves inside tissue phantom. The thermal wave speed is dependent on the thermal relaxation time, τ. Even though its value has been measured for biological media such as bologna meat samples [1], the exact value of thermal relaxation time (typically having values in the range of 5-100 sec) is known for most of the tissues. For this simulation result, the thermal relaxation time is selected as a 17 sec.

In Figure 4, the parametric study of thermal relaxation time is conducted. As the thermal relaxation time decreases, the temperature profile along the radial surface region seems not match with the experimental result. Especially, the parabolic model quite deviates and is hard to apply to the bio-heat transfer problem.

4. Conclusion

The hyperbolic conduction model for the bio-heat transfer problem was introduced. The three layered tissue with inhomogeneity was modeled and the 10 ps laser was deposited to ablate the inhomogeneity. The maximum temperature increased to about 65° C. The hyperbolic model was well matched to the experimental data rather than parabolic model. The non uniform grid system was employed to increase the calculation efficiency. The numerical result with non uniform grid system is more accurate without any drawback of calculation cost.

Table 1 Optical properties of tissues

Layer	Thickness (mm)	Absorption Coef.(mm^{-1})	Scattering Coef.(mm^{-1})
Epidermis	0.05	0.355	0.824
Dermis	3	0.049	0.824
Fatty tissue	10	0.05	0.55
Inhomogeneity	8	0.051	1.228

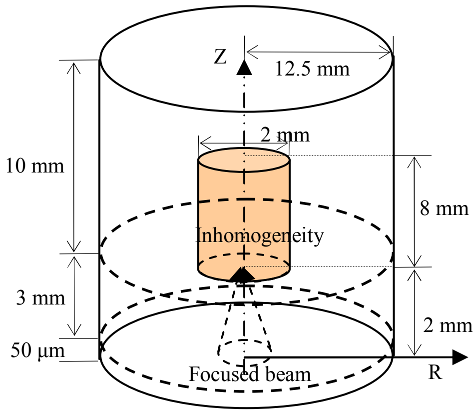

Fig. 1 The layered tissue model deposited with focused laser beam

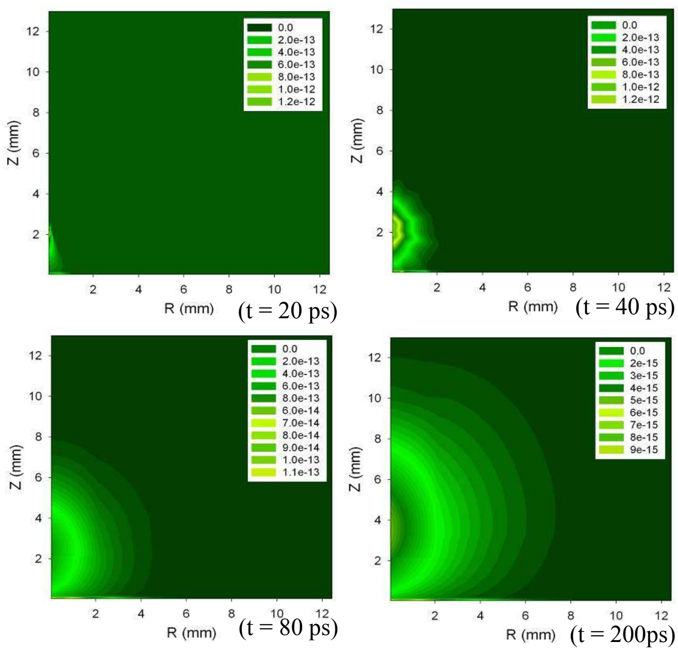

Fig. 2 The contours of divergence of radiative heat flux at selected time instants.

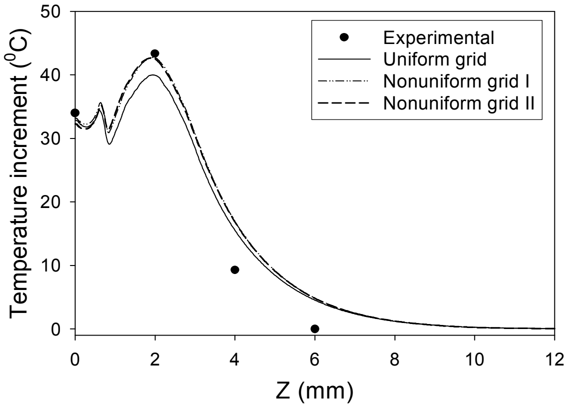

Fig. 3 The temperature profile along the z-axis at t = 10 sec.

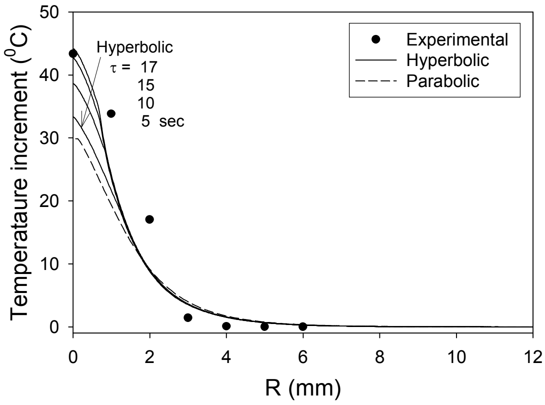

Fig. 4 The temperature profile along the radial surface at t = 10 sec.

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