초단파 레이저 조사시 티슈 열완화 시간 분석

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Analysis of Thermal Relaxation Time of Tissues Subject to Pulsed Laser Irradiation

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ABSTRACT

Two methodologies for predicting thermal relaxation time of tissue subjected to pulsed laser irradiation is introduced by the calculation the optical penetration depth and by the investigation of the temperature diffusion behavior. First approach is that both x-axial and y-axial thermal relaxation times are predicted and they are superposed to achieve the thermal relaxation time (τ_1) for two-dimensional square tissue model. Another approach to achieve thermal relaxation time (τ_2) is measuring the time required for local temperature drop until e^{-1} of the maximum laser induced heating.

Key Words: Thermal relaxation time, OPD (optical penetration depth), TDOM (transient discrete ordinate method)

1. Introduction

Since the first report on laser radiation,[1] many practical and potential applications have been investigated. Among them, medical laser surgery - a special laser material processing - certainly belongs to the most significant advances of the past several decades. Nowadays, clinical laser is widely used such as in optical tomography,[2,3] laser tissue ablation,[4] laser surgery,[5] and photodynamic therapy,[6] etc. Laser is also the excitation source in fluorescence imaging, scanning and labeling.[7] Fundamental to these applications is laser radiation transport associated with heat transfer in biological tissues. The general purpose of laser surgery among these applications is to provide proper heat to a targeted region of tissue. Furthermore, the challenging goal is to minimize the collateral thermal damage to the surrounding tissue of targeted region. The concept of thermal relaxation time is populated by selecting the pulse duration to minimize collateral thermal damage in microsurgery.[8] In this theory, the thermal relaxation time is mainly characterized by the tissue size and geometrics. If the tissue size is enough large-strictly speaking, larger than the optical penetration depth (δ), the thermal relaxation time (τ_{1x}) in one dimensional model can be predicted by

$$\tau_{1x} = \frac{\delta^2}{4\alpha} \tag{1}$$

where δ is the optical penetration depth and α

is the thermal diffusivity. The optical penetration depth can be defined that the fluence rate drops to e^{-1} of the original irradiance.[9] Furzikov used the beam diameter for δ to calculate the thermal relaxation time in lateral conduction matter.[10] However, tissue geometricsis the multi dimensional one and the spatial variance and size of incident laser can cause different thermal relaxation time. In present study, the two-dimensional square tissue model is prepared. For the x-axial thermal relaxation time (τ_{1x}), the optical penetration depth is measured. For the y-axial thermal relaxation time (τ_{1y}), the spatial fluence variation is used. Then, the thermal relaxation time (τ_1) is predicted by the superposing of τ_{1x} and τ_{1y}. The advantage of this method to predict thermal relaxation time (τ_1) is that the thermal relaxation time is only dependent on tissue and laser parameters. This is the approximate method but is good to prior information before laser surgery, which can be applicable gentle laser heating condition without forced convection or cooling.

To measure the optical penetration depth (δ), the fluence distribution induced by the laser light source inside tissue medium is required. If the medium exposed laser light source is dominant of absorbing characteristics, the optical penetration depth (δ) can be simply approximated as $1/\sigma_a$,[10,11] where, σ_a is an absorption coefficient. In case of absorbing and scattering medium (turbid medium), the σ_a can be replaced by σ_e (extinction coefficient) which is summation of the absorption and scattering coefficients. However, the fluence variation along the optical axis by this relationship was proved to deviate from fluence variation calculated by Monte Carlo simulation.[12] In current study, the fluence distribution is calculated by the TDOM (Transient Discrete Ordinate Method). Different with Monte Carlo method, it is the deterministic method to solve the transient radiative heat transfer problem. It is found that the TDOM can give satisfactory results for modeling laser radiation transport in turbid medium through previous research papers.[13,15] The algorithm of the TDOM provides not only light propagation behavior at certain time instant but the accumulated absorbing laser energy during laser pulse propagation. The 10^4 pulsed laser is generated like a pulse train until 1 ms and the accumulated absorbing laser energy is used to calculate the optical penetration depth (δ). Then, the thermal relaxation time (τ_1) of first introduced method is predicted with spatial thermal relaxation time (τ_{1y}).

Another approach to predict the thermal relaxation time (τ_2) is to measure the requiring time when the peak temperature rise to decrease to e^{-1} of the total rise in targeting region.[11] This thermal relaxation time (τ_2) is fully investigated by the transient heat conduction behavior. The temperature response during laser radiation transport is the consistent increasing without any heat conduction and convection. After eliminating laser light source, the tissue experiences the thermal relaxation process. The speed of the thermal relaxation is influenced by both optical and thermal properties. The tissue optical properties such as absorption and scattering coefficients causes to varying fluence distribution inside medium, which provides the initial condition of the heat diffusion process. Also, the thermal diffusivity, density, and heat capacity of tissue affects the thermal relaxation time. Typically, in this method, the convective heat transfer between air-tissue interfaces is incorporated. Thus, the thermal relaxation time (τ_2) is influenced by the convective boundary condition. This method seems to be more accurate compared with first method, the thermal relaxation time should be influenced by the environmental condition.

The objective of this research is threefold: (1) To investigate the methodology of measuring thermal relaxation time. Two methods are compared. (2) To investigate the parameters to affect thermal relaxation time. The optical properties (absorption and scattering coefficients), the beam width of deposited laser, and the convective heat transfer coefficient are considered. (3) To investigate accurate fluence distribution to measure optical penetration depth and spatial variation by the TDOM, temperature response during laser heating and thermal diffusion process.

2. MATHEMATICAL MODEL

2.1 Governing equation

Consider a collimated laser pulse incidence upon a 2-D biological tissue shown in Fig. 1. The local temperature response of the tissue during 10 ps pulse laser irradiation can be simply expressed as

$$\rho C_p \frac{\partial T(x,y,t)}{\partial t} = \nabla \cdot q_{rad}(x,y,t) \tag{2}$$

where ρ is the density, C_p is the specific heat, and $\nabla \cdot q_{rad}(x,y,t)$ is the divergence of radiative heat flux due to radiation absorption that is calculated by

$$\nabla \cdot q_{rad}(x,y,t) = \sigma_a (4\pi I_b - G) \tag{3}$$

where σ_a is the absorption coefficient of the tissue and I_b is the black body emissive power which is negligible because the tissue can be treated as a cold medium as compared to the large flux of laser beam. The incident radiation, G, is a direction-integrated radiation intensity and can be obtained by the summation of angle-discretized radiation intensity. To calculate the radiative intensity I^l in a discrete ordinate, time-dependent equation of radiative transfer (ERT) in discrete-ordinate format is introduced:

$$\frac{1}{c}\frac{\partial I^l}{\partial t} + \xi^l \frac{\partial I^l}{\partial x} + \eta^l \frac{\partial I^l}{\partial y} + \sigma_e I^l = \sigma_e S^l, \quad l = 1,2,3,\ldots,n \tag{4}$$

where ξ^l and η^l are the directional cosine in a discrete ordinate direction, σ_e is the extinction coefficient that is the sum of the absorption and scattering coefficients, and S^l is the radiative source term from the laser radiation. It can be expressed as

$$S^l = (1-\omega)I_b + \frac{\omega}{4\pi}\sum_{i=1}^{n} w^i \Phi^{il} I^i + S_c^l, \quad l = 1,2,3,\ldots,n \tag{5}$$

Here, the scattering albedo is $\omega = \sigma_s/\sigma_e$ and the scattering phase function is $\Phi(\hat{s}^i \to \hat{s}^l)$. A quadrature set of n discrete ordinate with the appropriate angular weight w^l $(l = 1,2,..,n)$ is used. The laser source S_c^l in Eq. (5) is the driving force of the transient radiation heat transport and can be expressed as

$$S_c^l = \frac{\omega}{4\pi} I_c (\mu^c \mu^l + \eta^c \eta^l + \xi^c \xi^l) \tag{6}$$

where the unit vector of (μ^c, η^c, ξ^c) represents the collimated laser incident direction. The incident laser sheet with spatially Gaussian profile can be expressed by

$$I_c(x,y) = I_{c0} \exp\{-4\ln 2 \times [(t - x/c)/t_p - 1.5]^2\} \times \exp[-(y - D/2)^2/v^2] \times \exp(-\sigma_e x) \quad (7)$$

in which I_{c0} is the peak amplitude of the intensity, which is modified to maintain 65°C of the maximum temperature rise. For example, I_{c0} is set up as 9.31×10^{-6} J/mm^2 in a fully scattering medium tissue with absorbing coefficient, $\sigma_a = 0.1mm^{-1}$. The t_p is the pulse width, D is the beam width, and v is the spatial variation factor. The beam width for this calculation is 2.5mm unless it specified. The Gaussian spatial ratio v/D is chosen as 0.25.

The temperature is nondimensionalized with its maximum temperature rising as,

$$\theta = \frac{T - T_i}{T_{ref} - T_i} \quad (8)$$

where, T_i is the initial tissue temperature and T_{ref} can be calculated in fully absorption medium as,

$$T_{ref} = T_i + \frac{\int_0^{\partial t_p} N_p \sigma_a I_c(x=0, y=D/2, t)dt}{\rho C_p} \quad (9)$$

where, N_p is the number of the laser pulse.

The laser source effect is diminished until 1 ms and the medium experiences heat diffusion from the irradiation region to the surrounding region without heat generation. This procedure is governed by the heat conduction equation by (10),

$$\rho C_p \frac{\partial T(x,y,t)}{\partial t} = k\left(\frac{\partial^2 T}{\partial x^2} + \frac{\partial^2 T}{\partial y^2}\right) \quad (10)$$

where k is the heat conductivity and the initial condition for this conduction equation is the temperature distribution inside of medium at 1 ms.

2.2 Boundary conditions

For radiation heat transfer, reflection and refraction governed by Snell's law and Fresnel equation, respectively, are considered at the tissue-air interface. At the rest boundaries, diffuse reflections are considered. For details, please refer to previous publications.[14,15]

For heat diffusion by the conduction after the irradiation source disappeared, the boundary conductions are specified below.

$$T(x,y,t) = 310\,(k) \quad \text{for } y = 0 \text{ or } y = W \quad (11\text{-}1)$$

$$T(x,y,t) = 310\,(k) \quad \text{for } x = L \quad (11\text{-}2)$$

$$Q_x = h(T_s - T_\infty) \quad \text{for } x = 0 \quad (11\text{-}3)$$

where T_∞ is the ambient coefficient (23°C) and h is the heat transfer coefficient.

2.3 The methodology to get thermal relaxation time

For two-dimensional square shape tissue medium, the x-axial directional thermal relaxation (τ_{1x}) and y-axial directional thermal relaxation time (τ_{1y}) are considered. Then, the thermal relaxation time (τ_1) is approximated as,

$$\frac{2}{\tau_1} = \frac{1}{\tau_{1x}} + \frac{1}{\tau_{1y}} \quad (12)$$

where, τ_{1x} can be calculated by Eq. (1) with a calculated OPD (δ) through numerical result and τ_{1y} can be achieved as,

$$\tau_{1y} = \frac{w_L^2}{2\alpha} \quad (13)$$

where, w_L is the spatial position in which original fluence peak drops to e^{-1}.

The second method to get thermal relaxation time (τ_2) is measuring transient temperature response during thermal diffusion process. In other words, the required time in which the nondimensional temperature reaches to e^{-1}.

2.2 Numerical Schemes

To solve the time-dependent equation of the radiative transfer, the TDOM is employed. The tissue geometry is divided by a uniform grid system of 200×200. The solid angle is divided by a quadrature set of $M = N(N+2)$ discrete ordinates for S_N method. In the present study, we use the S_{10} scheme. Details of the numerical schemes have been described in our recent publications[13-15]; thus, they are not repeated here. The fully explicit scheme is adopted to solve the heat conduction equation. The grid size is chosen same as radiative transfer problem and the time step is selected as 0.001 sec.

3. Results and discussion

The sketch of the tissue geometry is shown in Fig. 1. For simplicity, we considered a square tissue (L = W = 13.416 mm). The optical properties are chosen as selective way or actual human organ tissue: Selective σ_a (absorption coefficient) and σ_s' (reduced scattering coefficient) are chosen to investigate property effect and later the varying human tissue optical properties are adopted.[16] The thermal properties are assumed as the tissue is composed with 70% water.[11] The thermal diffusivity is α = 0.142 mm^2/s, and $\rho C_p = 3.52 \times 10^6 \mathrm{J/(m^3 \cdot K)}$. The width of incident laser sheet is 2.5 mm. This beam half width is approximated that the fluence drops to the e^{-2} of the original peak radiance.[11]

In Figure 2, the fluence variation along the optical axis is shown at 1 ms time instant. To validate code accuracy, the fluence profiles of the fully absorbing medium ($\sigma_a = 0.1 mm^{-1}$ and $\sigma_a = 1.0 mm^{-1}$) is compared with the analytical one. These graphs decay the exactly exponential behavior with the gradient of the absorption property. If the highly scattering medium is chosen like as $\sigma_s' = 1.3\, mm^{-1}$ or $\sigma_s' = 2.6\, mm^{-1}$, the fluence profiles drop quickly as scattering characteristics is enhanced. Here, the optical penetration depth (δ) is defined of the optical zone as the depth which the fluence has dropped to e^{-1} of the original radiance. For example, the optical penetration depth (δ) of the fully absorption medium with $\sigma_a = 0.1\, mm^{-1}$ is shown as 10 mm. However, the scattering medium with $\sigma_s' = 1.3\, mm^{-1}$, δ is dropped to the value less than 2 mm. If higher scattering medium $\sigma_s' = 2.6\, mm^{-1}$ is considered, δ becomes smaller, which is about 1mm. It is the 10 times smaller value of the fully absorbing medium with the $\sigma_a = 0.1\, mm^{-1}$. To predict the thermal relaxation time (τ_1), the accurate optical penetration depth measuring is required and the inverse relationship of absorption coefficient shows the quite deviate result when the highly scattering effect is contributed.

In figure 3, the y-axial beam variance is investigated with a normalized fluence value. The fluence distribution is shown with the 2.5 mm Gaussian beam width, which is approximated to the fluence by the e^{-2} drop of original fluence. The fluence distribution of the laser beam and fully absorbing case at 1 ms is well matched since the photon scattering does not exist. When the scattering medium is introduced, the width of the fluence distribution becomes wide. Here, we define another parameter, w_L. This is a y-axial fluence drop until

e^{-1} of the peak fluence similarly optical penetration depth.

The x-axial thermal relaxation time (τ_{1x}) using optical penetration depth and y-axial thermal relaxation time (τ_{1y}) using beam spatial variation are depicted varying optical properties in Fig. 4. The graphs shows that the τ_{1x} is changed highly as increasing absorption coefficient compared with the τ_{1y}. Since the optical penetration depth shows higher variation by changing optical properties. As investigated in Fig. 3, the spatial variation of the fluence in different optical properties is not quite large if the laser beam width is fixed. The fitted curves of the τ_{1y} varying scattering coefficients show the reverse order of those of the τ_{1x}. For example, the τ_{1y} of the fully absorbing medium is the shortest but the τ_{1x} of it shows the highest value. Both τ_{1x} and τ_{1y} converge to the corresponding thermal relaxation time if the absorbing characteristics is enhanced independent with the scattering coefficient. In higher absorbing medium, the scattering event is no more significant role to predict thermal relaxation time.

So far, we analyzed the thermal relaxation time with fluence distribution only considered radiation transport. Now, we introduce another method to predict the thermal relaxation time (τ_2) investigating thermal relaxation behavior. This method is the measuring local temperature response as time changing and thermal relaxation time is predicted by the e^{-1} drop of the maximum temperature rising. Exposing to air one side of the tissue phantom, the convective heat transfer should be considered. The temperature of the air is assumed to be the ambient temperature (23°C). The convective heat transfer coefficient (*h*) is selected as a $20W/m^2K$ to predict the thermal relaxation time (τ_2), which is followed by empirical result.[17] In many previous cases, the natural convective heat transfer coefficient for the tissue air interface is gained as a similar range. To investigate the influence of convective heat transfer coefficient (*h*), several h are compared in Fig. 5. The dermis tissue and heart tissue (endocardium) are compared. The optical properties are selected shown in figure. The thermal diffusivity of the heart tissue (endocardium) is chosen as α = 0.14 mm^2/s and $\rho C_p = 3.52 \times 10^6$ J/(m$^3 \cdot$ K). Obviously, the thermal relaxation time with higher *h* shows the small value both tissue models. It can be explained that larger convective heat transfer rate provides the rapid cooling performance. Thus, the local temperature is quickly dropped down and thermal relaxation time is predicted short value. Author points out one assumption that the convective heat transfer is negligible during laser irradiation. If convective heat transfer occurs together with laser radiation transfer, the relationship between *h* and τ_2 may be changed. The thermal relaxation of heart tissue (endocardium) is slow and the τ_2 turned out longer value due to the longer optical extinction distance. Furthermore, the influence of the h is significant to determine of the τ_2.

4. Conclusion

The temperature response upon laser irradiation has been investigated. The temperature mainly increased by the irradiation source and the thermal diffusion process followed by the heat conduction model. The thermal relaxation time is predicted by two methods. One is the measuring optical penetration depth and spatial fluence distribution. Then, both directional thermal relaxation times are calculated and superposed for two-dimensional square shape model. The tissue optical properties affect strongly the optical penetration depth. Both higher absorbing and scattering tissue has the short penetration depth and thermal relaxation time. If the absorption coefficient is over $\sigma_a = 2.0\,mm^{-1}$, the optical penetration depth became quite small independent with the scattering characteristics. The advantage of this approach is to predict thermal relaxation time only considering fluence distribution and any heat conductive transfer model is not required. However, this method shows the deviation result if the highly boundary condition effect is incorporated. Since, this method is only considered by the conductive heat transfer mechanism.

Another method of measuring local temperature response can predict more precisely if convective heat transfer is considered in tissue-air interface. Thermal relaxation time by both methods shows very similar aspect. However, this method is influenced by the convective heat transfer coefficient (h). With the larger beam width, the medium experienced slower rate of the heat diffusion and thermal relaxation time became longer. This is explained by the y-axial thermal relaxation time. As beam size increasing, the y-axial thermal relaxation time also increases. Thus, the thermal relaxation time of medium shows the longer one.

This calculated thermal relaxation time can provide good reference to select the laser pulse width. The Choi is proposed that the proper pulse duration time should be selected as a $1/100^{th}$ of the thermal relaxation time.[11] As Gaussian beam becomes shorter, the shorter pulse duration laser should be chosen due to the short thermal relaxation time.

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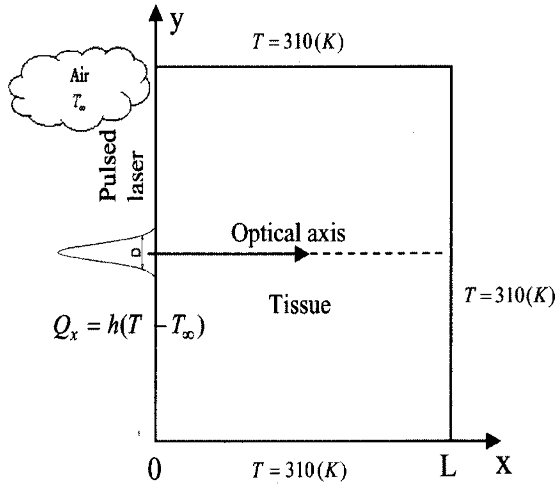

Fig. 1 Schematic diagram of laser tissue irradiation.

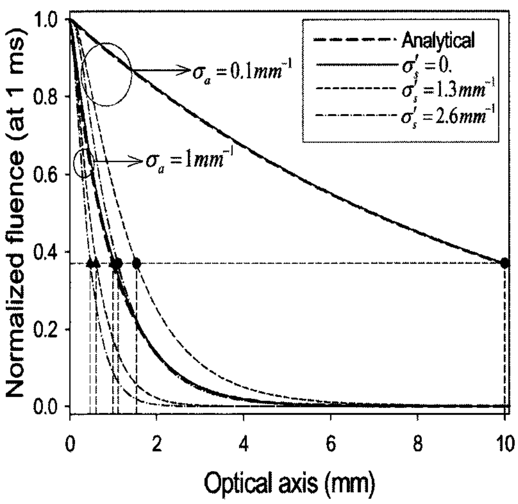

Fig. 2 The fluence variation along the optical axis at 1 ms for various optical properties.

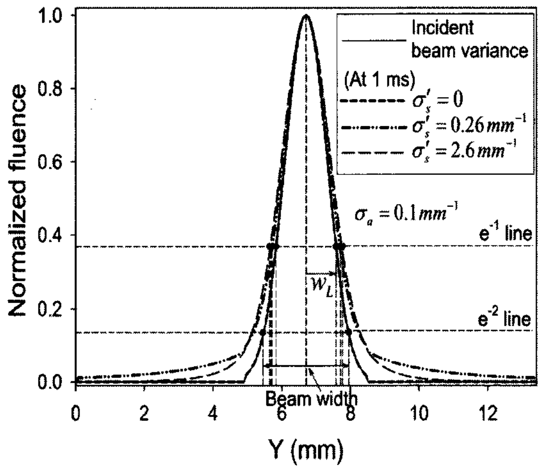

Fig. 3 The y-axial variance of the fluence with the different scattering coefficient conditions.

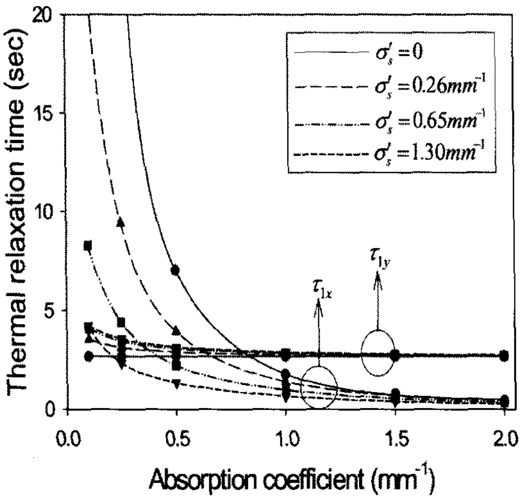

Fig. 4 The x-axial thermal relaxation time and y-axial thermal relaxation time with varying optical properties.

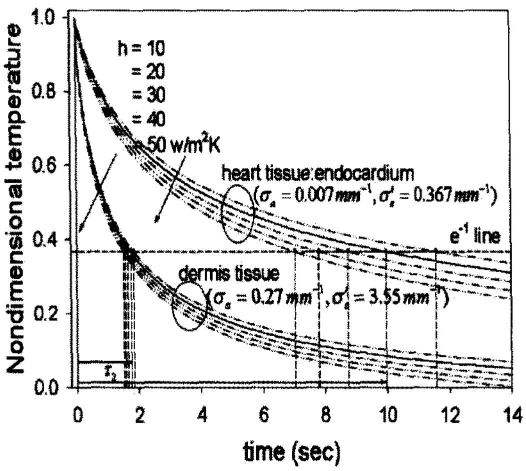

Fig. 5 The influence of the convective heat transfer coefficient to the thermal relaxation (τ_2), dermis and heart (endocardium) tissues are compared.