Photothermal damage prediction of laser interstitial thermotherapy

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An improved scattering optical model was developed under cylindrical coordinate to simulate the thermal effect of diffusing applicator in laser interstitial thermotherapy (LITT). The thermal damage was calculated by finite element method (FEM) using Pennes bio-heat transfer equation and Arrhenius injury integral formula. The numerical results showed that the scattering can considerably influence the evaluation of the lesion area, and the relationship between application powers or time and resulting tissue thermal damage was nonlinear. Although usually applying relatively low power can avoid tissue charring, rather higher power is recommended because it is indispensable to achieve necessary damage threshold and the therapy time can be shortened.

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Laser interstitial thermotherapy (LITT) is a recently developed minimally invasive technique aiming at destruction (coagulation) of solid small tumor by directly heating, while damage to the surrounding structures is strictly limited[1−4]. Infrared Nd:YAG laser (λ = 1064 nm) and diode laser (λ = 810 nm), which have deeper penetration depths, are usually used in LITT. Diffusing fiber applicators are used to minimize the potential carbonization (charring)[1,5]. The relationship between time/temperature exposure and resulting thermal damage is analysed for optimizing initial irradiation dosimetry and providing appropriate information on how to adjust the laser parameters and when to abort or stop the treatment[2,3,6−9]. A number of researches have aimed at modeling and understanding the thermal response of tissue during LITT with one-dimension (1D) simulation[3−5]. A two-dimensional (2D) theoretical investigations of lesion formation, scattering influence, power and time combination during single fiber LITT at 1064-nm irradiation in human liver are presented in this paper.

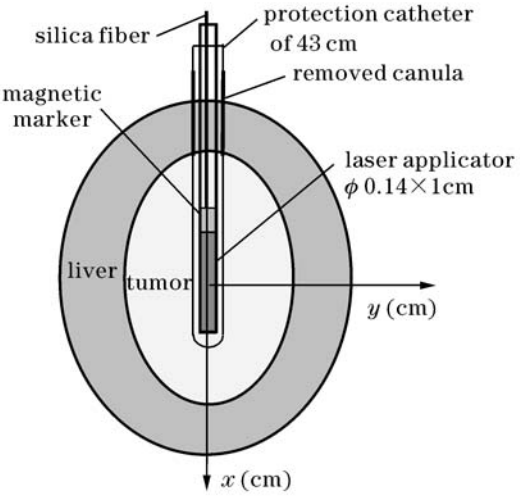

Fig. 1. Schematic diagram of laser coagulating tumor tissue.

The schematic diagram of LITT is shown in Fig. 1. The laser diffusing applicator was made by a homogeneous scattering glass coat mounted on a 400-μm silica fiber core. Laser light is transmitted to tissue with the diffusing applicator (with a diameter of 0.14 cm and effective distance of 1 cm). A protective catheter (with a length of 43 cm) prevents direct contact of the laser applicator with tissue. Under axis symmetry supposition, the temperature and damage distributions of half elliptical area can represent the whole volume, and the midpoint of the applicator is set as coordinate origin. Supposing the tumor has the shape of ellipsoid with axes of 1 and 2 cm. The computation area is also ellipsoid with axes of 2 and 3 cm. In the near infrared, laser interacts with tissue mainly by absorption and scattering, and the scattering dominates the absorption. Light diffusion approximation to the radiative transport equation is valid to model light transport in turbid biological tissue. The deposition of heat in tissue is due only to photons that are absorbed in the tissue and is modified by[7]

$$S = \begin{cases} \mu_{\rm a} I_0 \exp(-\mu_{\rm eff} y) \exp(-0.5\mu_{\rm s}' y), \\ \qquad\qquad |x| \le L \exp(0.5\mu_{\rm s}' y)/2 \\ 0, \qquad\quad \text{other} \end{cases}, \tag{1}$$

where $I_0 = P/2\pi r_0 L$ is the incident power intensity, r_0 and L are the applicator radius and effective length respectively, x and y are the axial and radial distances from the origin, respectively. According to beam spreading model when light transfers along the y axis, the light distribution will expand with a factor of $\exp(0.5\mu_{\rm s}' y)$, $\mu_{\rm s}' = (1-g)\mu_{\rm s}$ is the reduced scattering coefficient, $\mu_{\rm eff} = \sqrt{3\mu_{\rm a}[\mu_{\rm a} + (1-g)\mu_{\rm s}]}$ is the effective attenuation coefficient.

Because liver tissues are rather homogeneous, the classical Pennes bioheat transfer equation is used in the simulation as

$$\rho C \frac{\partial T}{\partial t} = \nabla(k\nabla T) + \rho_{\rm b} C_{\rm b} W_{\rm b}(T_{\rm b} - T) + S, \tag{2}$$

where T (K) is the temperature, t is the calculation time, S is the laser source, $W_{\rm b}$ (ml·g^{-1}·s^{-1}) is the blood per-

fusion rate, the subscript b means blood whose density is 1.06 g·cm^{-3} and heat capacity is 3.84 J·g^{-1}·K^{-1}[4]. The values of tissue optical, thermal, blood perfusion, and thermal damage parameters are listed in Table 1[3–5].

The *in vivo* initial temperatures and the liver boundary, which are far from the applicator, are defined as constant body core temperature of 37 °C. Another boundary at the cylinder axis is defined as adiabatic boundary because the net heat transflux through this surface is zero. The temperature distributions were calculated from Eqs. (1) and (2) by finite element method (FEM).

The evolution of the absolute temperature $T(r, z, t)$ at position (r, z) and time t was used to calculate the damage integral according to Arrhenius formula as

$$\Omega = \ln \frac{C_0}{C_t} = A \int_0^t \exp[-E/RT(r, z, t)]\mathrm{d}t, \qquad (3)$$

where C_t is the survival cell, C_0 is the original cell, R is the universal gas constant, A is the rate constant, E is the activation energy. According defining of first order damage, critical damage, and second order damage, Ω is 0.5, 1, 10, and there are about 60.7%, 36.7%, and 13.53% of C_0 cell still surviving respectively.

2D thermal response and thermal damage during laser irradiating in vivo human liver were simulated. The power intensity and irradiation time are taken into account to induce the determined damage. In Figs. 2—5 the application power is 10 W, the incident power intensity is 22.74 W/cm^2, and the irradiation time is 10 min.

In order to investigate the effects of scattering on the damage, we compare the calculations with scattering model and without scattering model (see Fig. 2). Critical damage results with or without scattering are shown. The ellipsoid damage shape of scattering model is more reasonable according to the experiment results[1].

The temperature rise through time at four points (y = 0.07 cm, x = 0, −0.5, −1, −1.5 cm) all get equilibrium after about 180 s because in vivo case blood will carry away part heat by perfusion (see Fig. 3), which plays a significant role in heat and mass transfer of biological tissue.

The damage contours Ω of 0.5, 1, and 10(see Fig. 4) are accordant to the temperature contours (see Fig. 5). When the two results are plotted on the same picture, the three damage contours will rest between the temperature contours of 50 and 60 °C which coincides with experiment of 53 °C[3]. The influences of power and irradiation time on first order damage (Ω = 0.5) dimensions (x and y) are shown in Fig. 6.

Table 1. Parameters Used for the Modeling of the Laser (λ = 1064 nm) and Human Liver Tissue Interaction

Parameter	Value
Perfusion Rate W_b	18.7×10^{-3} ml·g^{-1}·s^{-1}
Rate Constant A	1×10^{70} s^{-1}
Activation Energy E_a	4.0×10^5 J·mol^{-1}·K^{-1}
Absorption Coefficient μ_a	0.3 cm^{-1}
Scattering Coefficient μ_s	150 cm^{-1}
Anisotropy Factor g	0.93
Heat Conductivity k	5.66 mW·cm^{-1}·K^{-1}
Heat Capacity C	3.60 J·g^{-1}·K^{-1}
Tissue density ρ	1.05 g·cm^{-3}

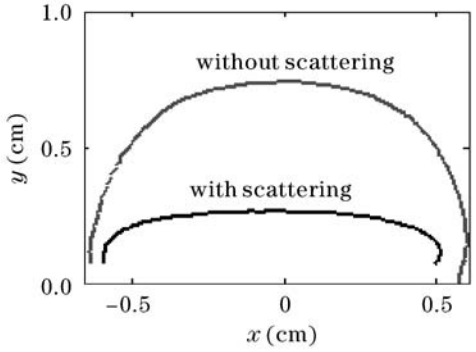

Fig. 2. Damage dimensions with and without scattering. P = 10 W, t = 600 s, Ω = 1.

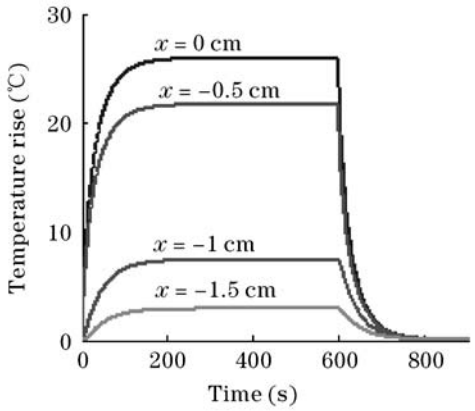

Fig. 3. Temperature rise versus time. P = 10 W, y = 0.07 cm.

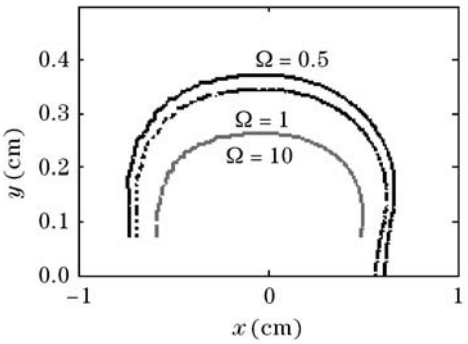

Fig. 4. Damage contour plot. P = 10 W, t = 600 s.

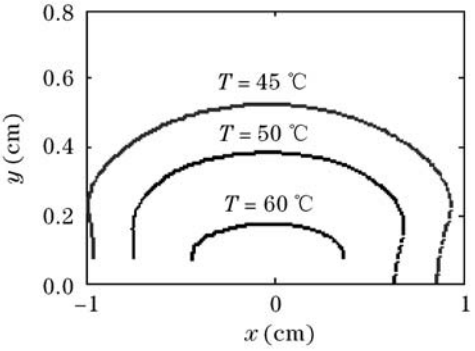

Fig. 5. Temperature contour plot. P = 10 W, t = 600 s.

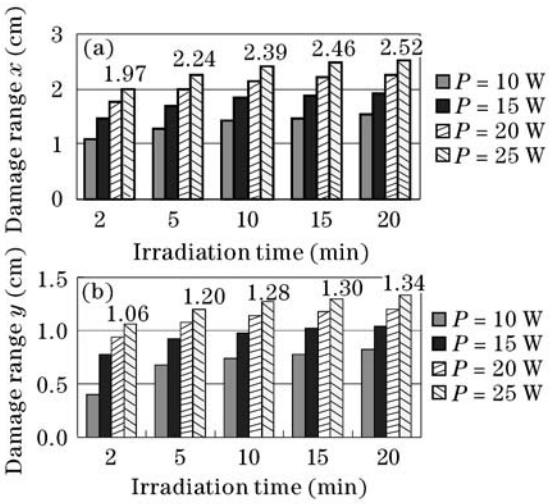

Fig. 6. Damage dimensions of different powers and irradiation times ($\Omega = 0.5$).

The dimension of x is about twice of that of y because the applicator is cylindrical. Increasing treating time will not increase the damage area dramatically because of the temperature equilibrium. If the temperature is rather low, it needs a rather long time (12 h at 45 °C) to get necessary damage to the tumor[9]. The damage depth of x increases only 0.32 cm from 5 to 20 min ($P = 25$ W) but increases 2.00 cm from 10 to 25 W ($t = 20$ min). So enough power intensity is indispensable to get high enough temperature and broader therapy dimension.

1D simulation was usually used in spherical applicator or naked fiber application but 2D simulation is more reasonable to LITT cylinder applicator. Dynamic optical parameters were not considered in this study because the coagulation was limited in small dimension even some researches showed that Arrhenius model was proved to be able to calculate the dynamic properties[5], the temperature will be lower and the dimension will be more expanded taking the increase scattering after coagulation into account.

Because human liver tissue has less absorption, higher thermal conductivity, and higher blood perfusion rate than those of beef, rat or pork, the temperature rise is lower than experiment results[10]. This study predicted that LITT is effective to an ellipsoid damage dimension of up to 2.52 cm in long axis and 1.34 cm in short axis at 25 W and 20 min irradiation. The necrosis size can be further expanded by multi-applicator technique. Long irradiation time is dispensable because at low power intensity the tumor will get low temperature equilibrium and the damage is not so easy to induce. Higher power intensity is suggested to achieve necessary injury to the tumor and the treating time will be shortened. In order to predict exact powers for different damage dimensions, the thermal damage inverse problem, namely, the detailed nonlinear relationship between damage dimension and application power and irradiation time waits for further research.

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