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A PRELIMINARY STUDY OF PULSED LASER HEATING OF TISSUE FOR THE IMPLICATIONS OF SKIN RESURFACING

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Abstract – Pulsed Er:YAG and CO_2 lasers induced temperature rise of tissue is studied using axisymmetric, two-dimensional, and transient Pennes' bio-heat equation for elucidating the implications of skin resurfacing. Modeling indicates that Er:YAG laser induced temperature has much higher but more shallow distribution in tissue than that of the CO_2 laser because of much higher absorption coefficient. The increase of repetition rate does not much affect on temperature rise because these laser modalities have much shorter heat diffusion time than the temporal length of each off-pulse. This model works as a tool to understand the photothermal effect in the laser-tissue interaction.

INTRODUCTION

Pulsed lasers in the mid and far infrared ranges are in clinical use for the treatment of facial rhytides and scars.[1-7] Resurfacing of the skin with a laser is a clinical procedure used to reduce facial rhytids,and to improve the over-all appearance and texture of facial skin. Mid-infrared Er:YAG laser has been recognized one of the best tool for drilling and cutting of hard tissues such as bone and tooth,[8-10] and its wavelength (2.94 μm) has the highest absorption peak in water which is the major constituent of the soft tissue.[11,12] An estimated laser penetration depth is on the order of 1 μm in cutaneous skin though actual ablation depth made is more deep than this value. As a result, this laser can be used for the treatment of aging facial skin and uneven pigmentation. Lower ablation threshold of the Er:YAG laser in tissue requires less laser energy to ablate the tissue so it minimized thermal damage around the ablated zone, and subsequently it insures fast and safe healing after clinical procedures.

Pulsed CO_2 laser has an increasing application for skin care and it also has strong application in the hard tissues treatment because the major constituent of hard tissue, hydroxypatite, has absorption peak near 10.6 μm wavelength.[13] Far-infrared CO_2 laser wavelength also has absorption peak in water.

The penetration depth of the CO_2 laser is more deep than that of the Er:YAG laser such that it yields thicker thermal damage zone around the ablated area.

This study has focused on the test of the axisymmetric, two dimensional, and transient Pennes' bio-heat equation in a semi-infinite medium employing two different laser systems which are common in the plastic and aesthetic surgery. This preliminary study will also be extended to the moving laser source system which is common in the articulated arm or fiber optic assisted system.

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METHODS

Let' s consider a pulsed, static laser beam irradiating the surface of a semi-infinite substrate.

The origin is on the surface (X-Y plane). The positive Z axis points into the substrate. The beam intensity has a Gaussian profile with a 1/e^2 radius. We assume that the laser beam is switched on for a finite time and then switched off. The rise and fall of the laser power are assumed instantaneous and we neglect heat loss from the surface. The substrate is assumed to have properties which are isotropic and temperature independent. Temperature rise in the tissue is calculated according to the Pennes' bio-heat equation that account for the presence of blood perfusion

$$\nabla^2 T - \frac{1}{\alpha}\frac{\partial T}{\partial t} = -\frac{Q}{\kappa} - \rho_b c_b \omega_b [T_\infty - T] \quad (1)$$

where $\alpha = \frac{\kappa}{\rho c}$ is the thermal diffusivity, and Q is the heat source term, ρ_b is the density of blood, c_b is the specific heat of blood, ω_b is the blood perfusion rate, and T_∞ is the environment temperature. The heat source term is

$$Q = \frac{2\mu_a P}{\pi \omega^2}\exp[-\frac{2(x^2+y^2)}{\omega^2}]\exp(-\mu_a z) \quad (2)$$

where μ_a is the absorption coefficient at the given wavelength, P is the laser power. In general the temperature rise at a point (X,Y,Z) can be obtained from the Green function for this problem:

Table 1. Thermophysical constants used in the model.[14]

μ_a(Er:YAG)	8000cm^{-1}(assumed 65% water content)
μ_a (Co_2)	500cm^{-1}(assumed 65% water content)
ρ_b	1.06 g/cm^3
c_b	3.84 J/g · k
ω_b	120 ml/kg · min
κ	0.096 w/cm · k
ρc	3.4 J/cm^3 · k
α	0.0016 cm^2/s (calculated using constants just above)
T_∞	25℃

$$G(x, y, z, t \mid x', y', z', t') = \exp\left[-\frac{\alpha\rho_b c_b\omega_b(t-t')}{\kappa}\right]\cdot\left[\frac{1}{4\pi\alpha(t-t')}\right]^{1.5}\cdot\exp\left[-\frac{(x-x')^2+(y-y')^2}{4\alpha(t-t')}\right]\cdot\left[\exp\left(-\frac{(z+z')^2}{4\alpha(t-t')}\right)+\exp\left(-\frac{(z-z')^2}{4\alpha(t-t')}\right)\right] \tag{3}$$

Combining Eqs. (2) and (3) produces a general solution during a single pulse laser heating:

$$T(x, y, z, t) = \int_0^t \frac{\mu_a P}{\pi\rho c_p}\exp[-\alpha\tau(\Gamma-\mu_a^2)]\cdot\exp\left[-2\frac{(x^2+y^2)}{\omega^2+8\alpha\tau}\right]\cdot\left[\exp(\mu_a z)\,\mathrm{erfc}\left(\frac{z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)+\exp(-\mu_a z)\,\mathrm{erfc}\left(\frac{-z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)\right]+\int_0^t \alpha\Gamma\exp(-\alpha\tau\Gamma)d\tau\cdot T_0 \tag{4}$$

where $\Gamma = \frac{\alpha\rho_b c_b\omega_b}{\kappa}$. This expression can be written in terms of cylindrical coordinate as follow:

$$T(r, z, t) = \int_0^t \frac{\mu_a P}{\pi\rho c_p}\exp[-\alpha\tau(\Gamma-\mu_a^2)]\cdot\exp\left(\frac{-2r^2}{\omega^2+8\alpha\tau}\right)\cdot\left[\exp(\mu_a z)\,\mathrm{erfc}\left(\frac{z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)+\exp(-\mu_a z)\,\mathrm{erfc}\left(\frac{-z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)\right]+\int_0^t \alpha\Gamma\exp(-\alpha\tau\Gamma)d\tau\cdot T_0 \tag{5}$$

where $\tau = t - t'$. The initial temperature T_0 is assumed zero for

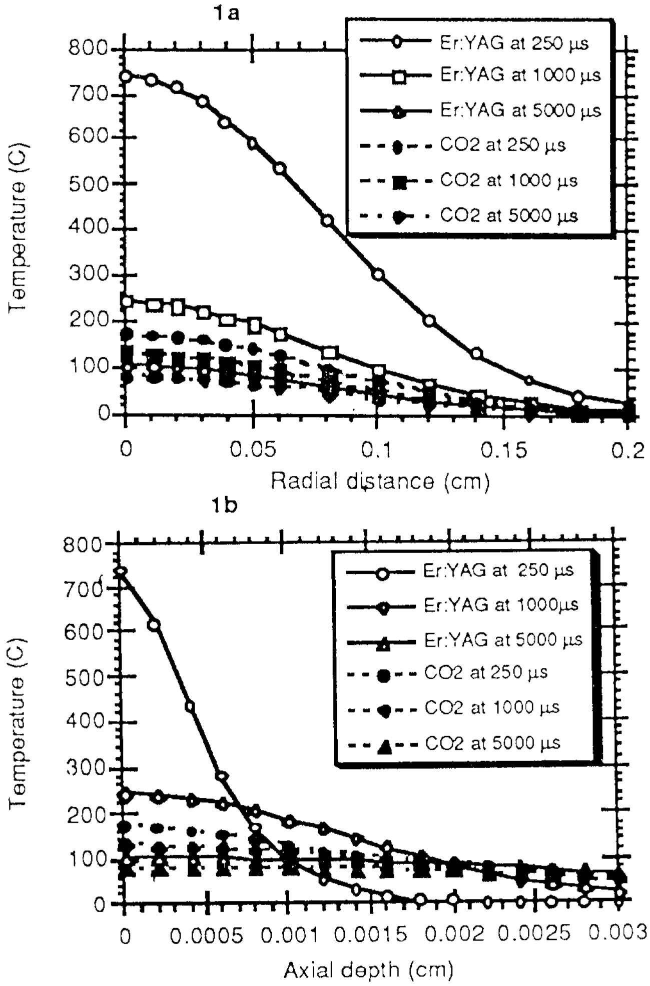

Figure 1. (a) Temperature distributions at radial distance. The laser power was 200 W. The beam radius was 0.15 cm.
(b) Temperature distributions at depth with the same power and beam radius as (a).

the convenience. However, when the time t falls in between two pulses (which corresponds to the cooling stage of the substrate), then

$$T(r, z, t) = \int_{t-\nabla t}^{t} \frac{\mu_a P}{\pi\rho c_p}\exp[-\alpha\tau(\Gamma-\mu_a^2)]\cdot\exp\left(\frac{-2r^2}{\omega^2+8\alpha\tau}\right)\cdot\left[\exp(\mu_a z)\,\mathrm{erfc}\left(\frac{z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)+\exp(-\mu_a z)\,\mathrm{erfc}\left(\frac{-z+2\alpha\mu_a\tau}{2\sqrt{\alpha\tau}}\right)\right]+\int_{t-\nabla t}^{t}\alpha\Gamma\exp(-\alpha\tau\Gamma)d\tau\cdot T_0 \tag{6}$$

where ∇t is the pulse width. The following table is the thermophysical constants used in the model.

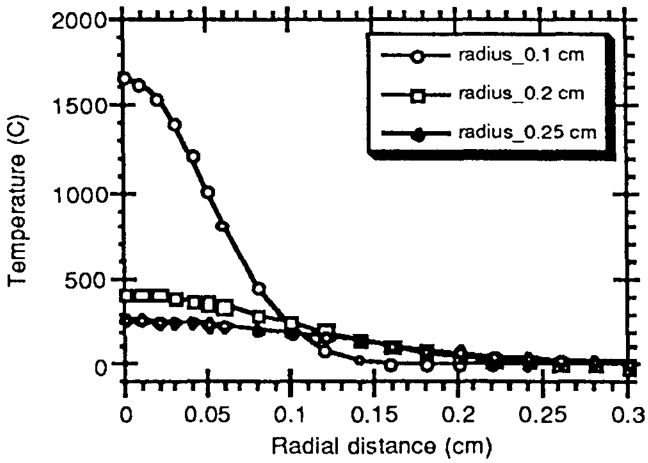

Figure 2. Effect of spot size on temperature distribution at the end of a single Er:YAG laser pulse. The laser power was 200 W.

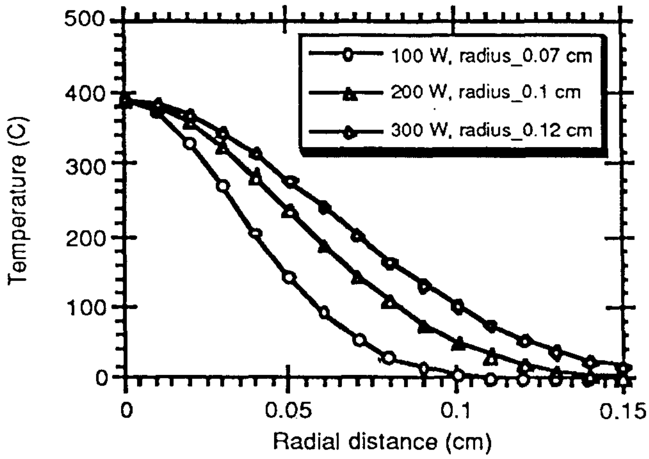

Figure 3. Effect of variation of fluence on thermal response. The power intensity and pulse width were 6300 W/cm^2 and 250 μs, respectively.

RESULTS AND DISCUSSION

The thermal response of human skin to different heating modalities, Er:YAG and CO_2 laser, has been studied for different pulse width, spot radius, power intensity, and pulse repetition rate. Temperature distribution for different lasers at and after the end of a single pulse are shown in Fig. 1. The pulse width was fixed to 250 μs for both lasers. Temperature rises due to the Er:YAG laser irradiation were much higher than those of the CO_2 laser because the absorption coefficient of 2.94 μm wavelength in water was much greater than that of 10.64 um wavelength. After one single pulse, we observed in Fig. 1(a) that the temperature was decreased smoothly on the surface because of heat diffusion to the substrate. Cooling rate of the Er:YAG laser in the radial direction was much higher than that of the CO_2 laser. The effect of blood perfusion in the temperature distribution, though not shown in the figure, was negligible because of its very low perfusion rate in the microsecond range.

Fig. 1(b) shows the temperature distribution at the substrate. The CO_2 laser allows deep heating in the tissue than that of the Er:YAG laser because of the deeper laser light penetration. The effect of heat diffusion is clear in the figure after one pulse such that the temperature at the substrate increases gradually as time goes.

The effect of laser beam size on temperature distributions at the end of a single Er:YAG laser pulse is shown in Fig. 2. For a fixed laser power, the temperature at the center becomes lower as the size of beam increases. However the gradient of temperature in the radial direction becomes less steep and relatively uniform temperature distribution becomes possible. Since the power intensity has inverse proportion to the laser irradiated area for a fixed laser power, the resultant thermal response has inverse proportion to the square of the given spot radius.

The effect of variation of fluence on thermal response is shown in Fig. 3. In the CO_2 laser heating, for the fixed power internsity (6300 W/cm^2) and pulse width (250 μs), higher laser power yields slower temperature decrease on the surface as it goes to the periphery though the increase of power and laser interaction area on the surface are same. It is clear from the figure that increasing the laser power with spot size is a better way to enhancing temperature rise on the surface than decreasing the power and spot size except for the center.

Figs. 4(a) and 4(b) show the effect of repetition rate on temperature rise at subsurface for different lasing system. The power of laser was fixed to 200 W per pulse. The ratio of temperature rise induced by the Er:YAG laser looks like slightly affected by the increase of repetition rate. The increase of temperature becomes higher as it goes deeper in the tissue and as the repetition rate increases. On the other hand, the ratio of temperature rise induced by the CO_2 laser looks like more uniform in the range we have investigated as the repetition rate increases.

It is believed that Er:YAG and CO_2 laser, though they have different wavelengths, have same ablation mechanism because of their common chromophore in the tissue: water. When the laser light is absorbed in water, temperature in the laser absorbed area increases tremendously within a very short time and which result in a pressure build-up. The built pressure induces explosive ablation and/or tissue tearing when it exceeds the ultimate strength of the tissue. The ejected particles exert backward force (pressure) onto the remaining tissue. Ablation and the secondary thermal damage is a combination of thermal and mechanical effects. Generally it is difficult to predict quantitatively the depth of ablation and thermal damage around the ablated zone because the ablation process accompanies

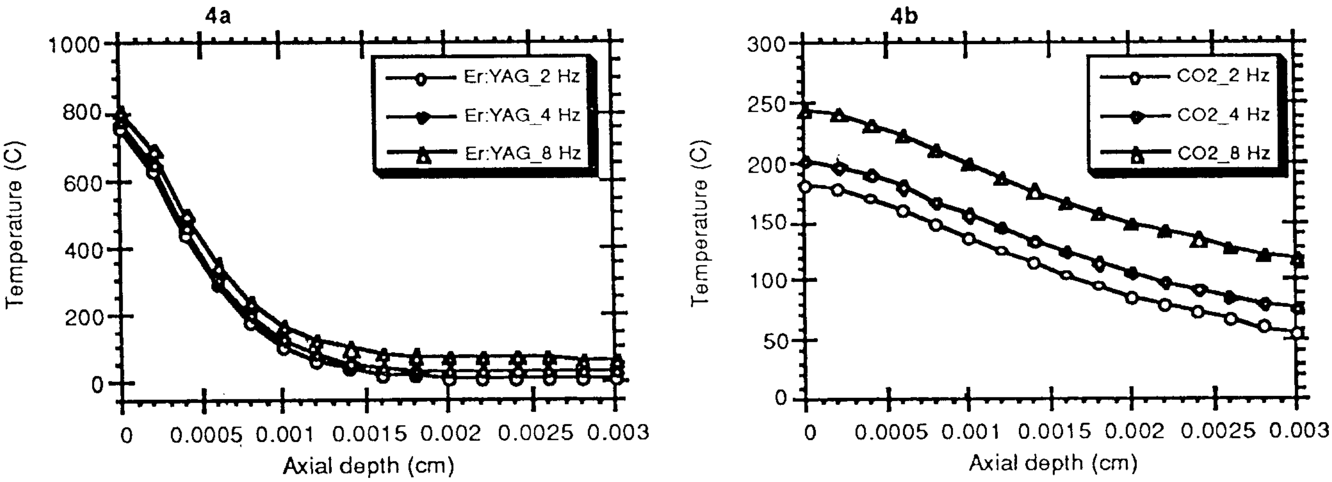

Figure 4. (a) Effect of repetition rate on temperature distributions due to the Er:YAG laser irradiation. (b) Effect due to the CO_2 laser irradiation.

dynamical changes of optical, thermal, and mechanical properties of the tissue. A complete model which could predict the ablation process by taking these dynamical change is not feasible yet. However, our simple model can provide meaningful predictions about the photothermal effects in the tissue and it is still important because this model allows the medical doctors to have indirect experience for the laser treatment of the skin through the proper parametric study.

The difference of the ratio of temperature decrease from 250 μs to 1000 μs for two different lasers can be understood by the heat diffusion in the substrate. The heat diffusion time in the laser irradiated zone depends on the penetration depth and thermal diffusivity of the tissue. The initiation of heat diffusion in the Er:YAG laser irradiated zone ($\sim$1 μs) is much faster than that of the CO_2 laser ($\sim$100 μs), and which result in greater temperature drop on the surface and subsurface.

The temporal distance between pulse to pulse is the primary factor which determine the rate of thermal energy accumulation in the laser irradiated zone. The initiation of heat diffusion in the soft tissue associated with the wavelength we have studied is in the microsecond range. On the other hand, the repetition rate we have chosen is so low and the temporal distance between pulse to pulse is in the millisecond range that the accumulation of thermal energy is minimal though the total input laser power was increased several times.

The effect of natural cooling of the laser irradiated tissue by the blood perfusion depends on the volume of blood and the perfusion time. The skin has very low blood perfusion rate than other internal organs (for example, kidney; male = 4000 mL/kg min, liver; male = 1000 mL/kg min, muscle; cardiac; male = 800 mL/kg min, etc.). The pulse width we have chosen is so short that the blood perfusion effect is negligible. Generally, cooling by air or water flow is more effective way. The advantage of cooling by convection is in the fact that we can easily control convection coefficient. The convection coefficient, h, depends on the thermal conductivity, and the thermal conductivity depends on the temperature such that by controlling the temperature of fluid we can handle the efficiency of cooling effectively.

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