# Pulsed Laser를 이용한 생체조직 가열을 위한 모델링: Skin Resurfacing을 위한 연관성

권 용 훈, S. Rastegar

# Modeling of Pulsed Laser Heating of Tissue: Implications for Skin Resurfacing

Y. H. Kwon, S. Rastegar
Biomedical Engineering Program, Texas A&M University,
College Station, TX 77843, U.S.A.

## Abstract

Pulsed Er:YAG and $CO_2$ lasers induced temperature rise of tissue are studied using axisymmetric, two-dimensional, and transient Pennes' bio-heat equation for the implications of skin resurfacing. Model results indicate that Er:YAG laser induced temperature has much higher but more shallow distribution in tissue than that of the $CO_2$ laser because of its higher absorption coefficient. The increase of repetition rate does not affect the temperature rise too much 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. [1,2] Resurfacing of the skin with a laser is a clinical procedure used to reduce facial rhytids, improve the over-all appearance and texture of facial skin, and to revise disfigurements such as scars. Mid-infrared Er:YAG laser has recognized one of the best tool for drilling and cutting of hard tissues such as bone and tooth [3] and its wavelength (2.94 μm) has the highest absorption peak in water which is the major constituent of the soft tissue [4]. 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. It minimizes thermal damage around the ablated zone, and subsequently it ensures fast and safe healing after clinical procedures.

Pulsed $CO_2$ laser has an increasing application in skin care. It also has strong application in the hard tissues treatment because the major constituent of hard tissue, hydroxyapatite, has absorption peak near 10.6 μm wavelength [5]. Far-infrared $CO_2$ laser wavelength also has absorption peak in water. The laser penetration depth of the $CO_2$ laser is more deep than that of the Er:YAG laser and so it yields a 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. It employs 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

## Theory

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/e2 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

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

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

$$Q = \frac{2\mu_a P}{\pi\omega^2}\exp\left(-\frac{2(x^2+y^2)}{\omega^2}\right)\exp(-\mu_a z), \qquad (2)$$

where $\mu_a$ 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:

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

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

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

This expression can be written in terms of cylindrical coordinates as follows

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

where $\tau = t - t'$. The initial temperature $T_0$ is assumed zero for convenience. However, when the time t falls in between two pulses ( which corresponds to the cooling of the substrate), then

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

where $\nabla t$ is the pulse width. The following table shows the thermophysical constants used in the model.

## Results

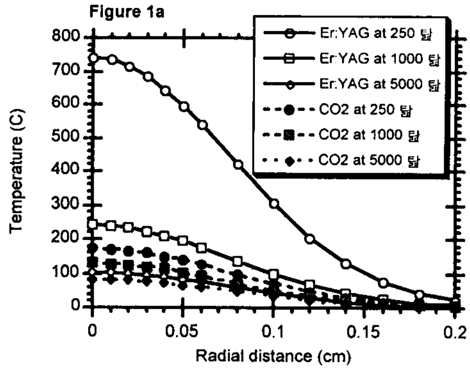
The thermal response of human skin to different heating modalities, Er:YAG and $CO_2$ laser, has been studied for different pulse width, spot radius, power intensity, and pulse repetition rate.

Table 1. Thermophysical constants used in the model [6].

| | |
|---|---|
| $\mu_a$ (Er:YAG) | 8000 $cm^{-1}$ (assumed 65 % water content) |
| $\mu_a$ ($CO_2$) | 500 $cm^{-1}$ (assumed 65 % water content) |
| $\rho_b$ | 1.06 $g/cm^3$ |
| $c_b$ | 3.84 $J/g \cdot k$ |
| $\omega_b$ | 120 $ml/kg \cdot min$ |
| $\kappa$ | 0.036 $w/cm \cdot k$ |
| $\rho c$ | 3.4 $J/cm^3 \cdot k$ |
| $\alpha$ | 0.0016 $cm^2/s$ (calculated using constants just above) |
| $T_\infty$ | 25 °C |

Temperature distributions for different lasers at and after the end of a single pulse are shown in Figure 1. The pulse width was fixed to 250 μs for both lasers. Temperature rises due to the Er:YAG laser were found to be much higher than those of the $CO_2$ laser because the absorption coefficient of 2.94 μm wavelength in water is much greater than that of 10.64 μm wavelength. After one single pulse, we observe in Figure 1(a) that the temperatures decrease smoothly on the surface because of heat diffusion to the substrate. Cooling rate of the Er:YAG laser in the radial direction is much higher than that of the $CO_2$ laser. The effect of blood perfusion in the temperature distribution, though not shown in the figure, is negligible because of its very low perfusion rate in the microsecond range. Figure 1(b) shows the temperature distribution at the subsurface. 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 when the temperature at the subsurface 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 Figure 2. For a fixed laser power, the temperature at the center becomes lower as the size of the



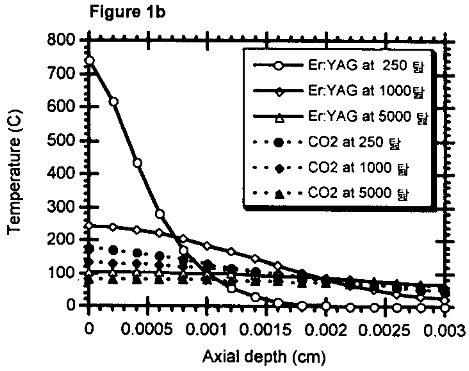




Figure 1. (a) Temperature distributions at radial distance. The laser power was 200 W. The beam radius was 0.15 cm. (b) Temperature distributions at depth with the same power and beam radius as (a).

beam increases. However the gradient of temperature in the radial direction becomes less steep and relatively uniform temperature distribution becomes possible. Since for a fixed laser power the power intensity has inverse proportion to the laser irradiated area, 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 Figure 3. For the fixed power intensity (6300 W/cm2) and pulse width (250 μs) in the $CO_2$ laser heating, 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.

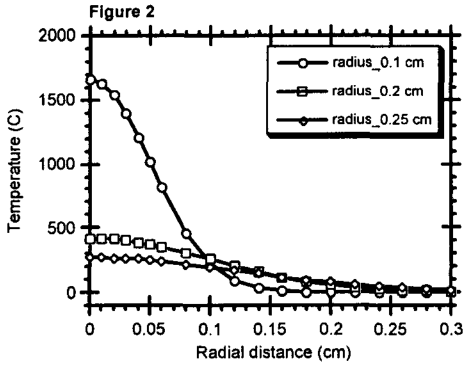



Figure 2. Effect of spot size on temperature distribtution at the end of a single Er:YAG laser pulse. The laser power was 200 W.

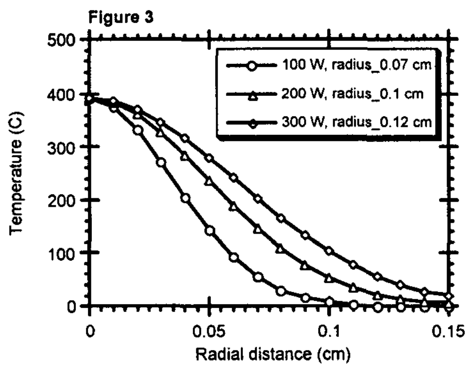
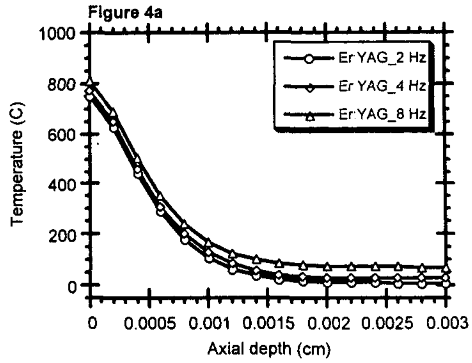



Figure 3. Effect of varaiation of fluence on thermal response. The power intensity and pulse width were 6300 W/cm2 and 250 μs, respectively.



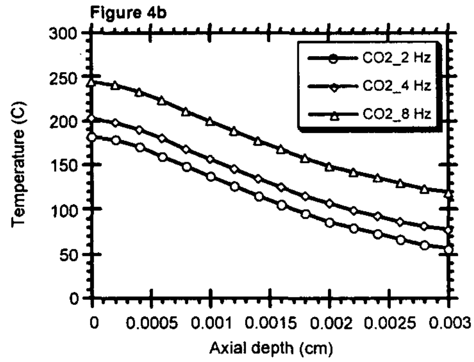




Figure 4. (a) Effect of repetition rate on temperature distributions due to the Er:YAG laser irradiation. (b) Effect due to the $CO_2$ laser irradiation.

Figure 4(a) and 4(b) show the effect of repetition rate on temperature rise at the subsurface for different lasing system. The power of the laser was fixed to 200 W per pulse. The temperature rise induced by the Er:YAG laser looks 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.

## Discussion

It is believed that though the Er:YAG and $CO_2$ lasers have different wavelengths they have the 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 this results 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 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.

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 (~1 μs) is much faster than that of the $CO_2$ laser ( ~100 μs), and which result in greater temperature drop on the surface and subsurface.

The temporal distance between two pulses is the primary factor which determines the rate of thermal energy accumulation in the laser irradiated zone. In the range of wavelengths that we used, the initiation of heat diffusion in the soft tissue was in the microsecond range. On the other hand, since the repetition rate we have chosen is so low and since the temporal distance between pulses is in the millisecond range, 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 a very low blood perfusion rate compared to 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 a more effective way. The advantage of cooling by convection is in the fact that we can easily control the 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 the fluid we can handle the efficiency of cooling effectively.

## References

1. R. Kaufmann, R. Hibst, "Pulsed Erbium:YAG laser ablation in cutaneous surgery", Lasers Surg Med 19, pp.324-330, 1996

2. R. E. Fitzpatrick, M. P. Goldman, "Advances in carbon dioxide laser surgery", Clinics Dermatol 13, pp. 35-47, 1995

3. R. Hibst, U. Keller, "Experimental studies of the application of the Er:YAG laser on dental hard substances: I. Measurement of the ablation rate", Lasers Surg Med, 9, pp. 338-344, 1989

4. G. M. Hale, M. R. Querry, "Optical constants of water in the 200-nm to 200 μm wavelength region", Appl Opt, 12, pp. 555-563, 1973

5. M. Forrer, M. Frenz, V. Romano, H. J. Altermatt, H. P. Weber, A. Silenok, M. Istomyn, V. I. Konov, "Bone-ablation mechanism using $CO_2$ lasers of different pulse duration and wavelength", Applied Physics B56, pp. 104-112, 1993

6. F. A. Duck, *Physical Properties of Tissue: A Comprehensive Reference Book, (Academic Press, 1990)*