Simulation of Temperature Distribution in Different Human Skin Types Exposed to Laser Irradiation with Different Wavelengths and Laser Irradiation Intensities

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Simulation of Temperature Distribution in Different Human Skin Types
Exposed to Laser Irradiation with Different Wavelengths and Laser Irradiation Intensities

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Abstract

In modern medical facilities, laser is becoming more important in aesthetic and medical applications. However, inappropriate laser irradiation can result in harmful effect such as thermal burn injury. This is owing to the thermal response of the skin due to laser irradiation is not well understood. In this study, the thermal response of skin to laser radiation was investigated under the effects of wavelength, laser intensity and incident time. The transient bioheat model was considered and solved numerically using a finite element method. This study illustrated that the skin temperature distribution depends strongly on the wavelength, intensity, and skin color type. The amount of energy absorb is significantly higher in the shorter wavelength range and with the increasing intensity. Further, the darker skin color result in higher laser absorption. The obtained values provide an indication of the limitations that must be considered during laser-induced thermotherapy.

Keywords: laser, bioheat equation, skin, finite element, thermotherapy
1. **Introduction**

Lasers have become extremely important treatment devices in the field of engineering and dermatology (Gould, 1959; Carroll & Humphreys, 2006; Maiman, 1960; Zaret et al., 1961). Recently, they have been used for variety of applications, ranging from surgical operations and the treatment of skin disease and cosmetic dermatology. In comparison to traditional methods, laser irradiation is more effective and reliable, featuring small wound and quick recovery (Carroll & Humphreys, 2006). In these treatments, the most important issue is to control the increase in temperature and distribution of temperature in the living tissue. This will lower the unwanted thermal damage to surrounding tissue. The thermal response of tissue to laser light depends on the thermal and optical properties, light source, duration of exposure, etc. In laser thermotherapy, prediction of safe dosage of laser light and irradiation time for tissue can be predicted accurately using numerical simulation techniques before its application for treatment (Carroll & Humphreys, 2006). Indeed, the Pennes bioheat equation is widely used for numerical simulation of laser thermotherapy. This equation was introduced by Pennes (1998), which is basically a heat diffusion equation incorporating the effect blood perfusion in capillaries and metabolic heat generation due to metabolism.

Recently, thermal models of biological tissue have been rapidly developed and have been used extensively in studies of dermatological implications during exposure to laser irradiation and prediction of therapeutic responses to laser irradiation. Dua and Chakraborty (2005) numerically investigated the damage to biological tissues when they were subjected to single-point laser diathermy. Jaunich, Raje, Kim, Mitraand, and Guo (2008) analyzed the temperature distributions and heat-affected zone in skin tissue when it was irradiated by a focused laser beam from a short pulse laser source. Lee and Lu (2004) carried out a numerical analysis of heat
transfer in a 3D skin tissue model with an embedded vascular system. In their study, the tissue temperature and the blood temperatures in arterial and venous vessels were solved by using the bioheat equation. Chen, Liang, Zhu, Sun, and Wang (2014) carried out a numerical calculation solution of the finite element method (FEM) of a skin tissue after laser irradiation with four wavelengths: 532, 694, 755, and 800 nm. Particularly, the change in temperature in perfused tissue due to diffusion of blood in the tissue and metabolic heat generation of any biological sample can be accurately predicted by Pennes' bioheat equation. However, this equation fails to predict correct temperature distribution in a vascularized tissue featuring large blood vessels. In such situation, the effect of blood flow in relatively large diameter vessels and on tissue temperature can be determined using combined momentum and energy conservation equations in the vascular domain and Pennes' bioheat equation in tissue domain. However, the present study consider a region of skin far from the large diameter blood vessels. The above assumption concurs with Bhowmik et. al. (2015) featuring the effect of focused laser light on perfused tissue and that of focused ultrasound on vascularized tissue. In fact, the rise in temperature and quantification of thermal damage in the skin and tumor due to focused laser light was determined by combined bioheat equation, laserlight attenuation and thermal damage equation.

However, most studies have mainly focused on the modeling and influence of specific parameters such as the wavelength. In fact, a systematic studies considering the effects of operating parameters such as laser wavelength, laser irradiation intensity, type of tissue, and irradiation time on the heat transfer in layered skin are scarce. These parameters directly affect the therapeutic effectiveness during the treatment. In practical situations, these effects enhance the heat transfer and absorption process within the target tissue, which can cause changes in temperature within the tissue. Therefore, in order to provide adequate information on the
appropriate level of laser transition from a laser instrument, it is essential to consider all of the previously mentioned parameters in the analysis.

The objective of this research is to model the laser–tissue interaction to contribute to medical applications, by understanding the thermal response of tissue to laser irradiation. In order to minimize unwanted side effects from the thermal damage and to provide a basis for new treatment strategies by proposing prediction and evaluation method by developing models and simulation tools. The effects of the laser wavelength, laser intensity, type of tissues, and irradiation time on heat transfer during laser-induced thermotherapy in the layered skin are systematically investigated.

2. Formulation of the Problem

According to the real biological structure, the skin is divided into three layers: epidermis, dermis, and subcutaneous tissue (Odland & Goldsmith, 1991). When the skin tissue is exposed to laser irradiation, the temperature gradient within the layers of the outer skin, epidermis and dermis, play an important role in the heat conduction. This study determines the effects of operating parameters on the heat transfer in the layers of skin during laser irradiation.

3. Methods and Model

The study focuses on the heat transfer characteristics of skin subjected to a laser beam from different therapeutic situations. The numerical simulation is analyzed by FEM on COMSOL Multiphysics software. The temperature changes in the layered skin due to the energy absorbed are described by Beer–Lambert's law and Pennes bioheat equation.

3.1 Physical Model
A 2D axisymmetrical model of the layered skin, which refers to the physical model in the previous research, is developed (Li, Li, Huang, & Xu, 2010). Figure 1 shows the 3D and 2D planes of the skin model used in this study. The skin model can be simplified to two-dimensional (2D) plane or axisymmetric analytical problems instead of three-dimensional (3D) simulation models in order to quick modeling and efficient analysis to meet the timing in computational simulation. Although the skin is a complex heterogeneous tissue, it is common to approximate skin as a 2D skin model constructed in an axial symmetrical plane with three different layers (epidermis, dermis, and subcutaneous tissue), as illustrated in Fig. 1(b). It supposes that the laser spot shape is circular and laser irradiated direction is perpendicular to the layered skin. It was also assumed that the tissue is approximately uniform in the same layer, which means that the thermal and optical properties do not different in the same layer of tissue. Table 1 gives the thicknesses, thermal and optical properties used in this study (Aguilar, Diaz, Lavernia, & Nelson, 2002; Bhowmik, Pepaka, Mishra, & Mitra, 2015; Chen, Liang, Zhu, Sun, & Wang, 2014; Tseng, Bargo, Durkin, & Kollias, 2009).

### 3.2 Mathematical Modeling

The temperature distribution within the layered skin is obtained by solving Pennes bioheat equation (Pennes, 1998). The transient bioheat equation effectively describes how heat transfer occurs within the skin, and the equation can be written as:

\[
\rho C \frac{\partial T}{\partial t} = \frac{d}{dz} \left( k \frac{dT}{dz} \right) + \frac{d}{dr} \left( k \frac{dT}{dr} \right) + \rho_b C_b \omega_b (T_b - T) + Q_{\text{met}} + Q_{\text{Laser}} \tag{1}
\]

where \( \rho \) is the tissue density (kg/m\(^3\)), \( C \) is the heat capacity of tissue (J/kg K), \( k \) is the thermal conductivity of tissue (W/m K), \( T \) is the tissue temperature (°C), \( T_b \) is the temperature of blood (°C), \( \rho_b \) is the density of blood (kg/m\(^3\)), \( C_b \) is the specific heat capacity of blood (J/kg K), \( \omega_b \) is
the blood perfusion rate (1/s), $Q_{\text{net}}$ is the metabolic heat generation (W/m$^3$), and $Q_{\text{Laser}}$ is the external heat source term related to laser irradiation (W/m$^3$).

In the analysis, heat convection between tissue and blood flow is approximated by the blood perfusion term, $\rho_b C_b \omega_b (T_b - T)$.

To simplify the problem, the following assumptions are made:

1. The layered skin tissue is a bio-material with constant thermal properties and optical parameters in the same layer.
2. There is no phase change of substance in the tissue.
3. There is no chemical reaction in the tissue.
4. 2D skin model constructed in an axial symmetrical plane is assumed.
5. Unsteady heat transfer is considered.
6. The contact surface between tissues is in a smooth condition.
7. The effect of shrinkage is negligible.
8. All tissues are homogeneous and isotropic.

The boundary conditions and the physical domain are indicated in Fig.2. The upper surface of the layered skin ($z = 0$) is considered to be under a convective boundary condition:

$$-kdT/dz = h_{am} (T - T_{am})$$

(2)

where $T_{am}$ is the ambient temperature (°C) and $h_{am}$ is the convective coefficient of the air (W/m$^2$ K).

The outer surface of the skin tissue, except for the upper surface, is considered to be under a constant core body temperature ($T_c$).
It is assumed that no contact resistance occurs between the internal layers of the three different layers (epidermis, dermis, and subcutaneous tissue). Therefore, the internal boundaries are assumed to be under a continuity boundary condition,

\[ n \cdot \left( k_i \nabla T_i - k_j \nabla T_j \right) = 0 \]  

where \( i \) is any layer and \( j \) is the adjacent layer to \( i \). Therefore, \( i = 1, 2 \) and \( j = 2, 3 \).

Considering the laser beam irradiation, the boundary condition will be applied on the boundary of the model in the direction of the laser beam in order to simplify the solution. Therefore, the laser intensity along tissue depth (\( z \)) is described by Beer–Lambert's law as follows:

\[ I (z) = I_0 e^{-az} \]  

The energy absorption of the laser irradiation can be expressed as follows:

\[ Q_{\text{Laser}} = aI_0e^{a\sigma^2} \]  

where \( I \) is the laser irradiation intensity (W/m\(^2\)), \( I_0 \) is the irradiation intensity at the skin surface (W/m\(^2\)), \( a \) is the absorptivity of the tissue (1/m), \( z \) is the depth of tissue, and \( \sigma \) is the width of the irradiated area (mm).

In this study, the absorption coefficient of the laser in the epidermis of the skin is taken from Tseng, Bargo, Durkin, and Kollias (2009), while the absorption coefficients of the dermis and subcutaneous fat are taken from Chen, Liang, Zhu, Sun, and Wang (2014) (see Table 1).

3.3 Numerical Procedure

The heat generation is determined by optical properties of tissue and laser parameters.
They are primarily the irradiance, irradiation time, and absorption coefficient, with this coefficient itself being a function of the laser wavelength. Heat transport is solely characterized by thermal properties of tissue such as heat conductivity and heat capacity. In addition, heating effects depend on the type of tissue and the temperature reached inside the tissue (Alster & Lupton, 2001; Fodor, Elman, & Ullmann, 2011). In this study, the bioheat equation and related boundary conditions are numerically simulated by using finite element solver, COMSOL Multiphysics software.

The 2D geometry is discretized using triangular elements. A convergence test of the wavelength (532 nm), intensity (1.5 W/mm²), and various properties listed in Table 1 is carried out to identify the number of elements required. The convergence test leads to a grid with approximately 40,000 elements. It is reasonable to confirm that, for this number of elements, the accuracy of the simulation results is independent from the number of elements.

4. Results and Discussion

In this study, the effects of laser light wavelength, incident time, laser intensity, and type of skin color on the temperature distributions in the layered skin during laser thermotherapy are systematically examined. Here, the mathematical model composed of the bioheat equation and Beer–Lambert’s law. These equations are solved simultaneously for different the therapeutic situations (Tung, Trujillo, López, Rivera, & Berjano, 2009; Cvetkovic, Poljak, & Peratta, 2010; Gheitaghy, Takabi, & Alizadeh, 2014). The following discussion focuses on the heat transfer in three-layered skin tissues during laser-induced thermotherapy (non-ablative laser irradiation). In this study, wavelengths of 532, 755, and 800 nm are selected. These wavelengths are referred from actual clinical procedures (Wall, 2007; Patil & Dhami, 2008; Trivedi, Yang, & Cho, 2017). It was assumed that the output of the laser is continuous. Further, laser irradiation intensities of
1, 1.5, and 2 W/mm$^2$ are selected. It is supposed that the laser beam is circular and that the laser irradiates the skin perpendicularly. In this study, the thicknesses of the epidermis, dermis, and subcutaneous fat are set as 0.05, 1.95, and 10 mm, respectively, and the radius of the domain is 10 mm.

Melanin functions as an absorptive pigment, which are ordinarily contained within the epidermis layer of the skin. It is an effective absorber of light and is able to select light color (wavelength) absorption (Patil & Dhami, 2008; Nouri, 2011). The absorption describes a reduction in energy and transform to thermal energy and finally result in increasing temperature. The size, shape and density of melanin particles is a factor in determining degree of opacity or skin color (Sturm, Box, & Ramsay, 1998). Darker skin contains more melanin (Yamaguchi, Brenner, & Hearing, 2007). Therefore, the darker skin can absorb more energy and generate higher temperature when compare to lighter skin. Consequently, the skin color is one selected parameter in this study. Referring to Fitzpatrick’s scale (Fitzpatrick, 1988), the skin color type is separated into six types: I, II, III, IV, V, and VI, denoting white, white-beige, light brown, moderate brown, dark brown, and black colors, respectively. The skin color types I–II and V–VI are chosen in this study as they can be clearly differentiated.

For the simulation, the optical properties and thermal properties are directly taken from Table 1. In all simulations, two skin types are selected, namely skin type I–II and skin type V–VI. We use their absorption coefficients of different wavelengths in the epidermis of the upper inner arm from Tseng, Bargo, Durkin, and Kollias (2009). Nevertheless, for all these laser wavelengths, the absorption coefficients of the dermis and subcutaneous tissues are taken from Aguilar, Diaz, Lavernia, and Nelson (2002) as 0.24 (1/mm) and 0.24 (1/mm), respectively. The thermal properties and skin thickness are taken from Chen, Liang, Zhu, Sun, and Wang (2014).
and the metabolic heat generation is taken from Bhowmik, Repaka, Mishra, and Mitra (2015), as shown in Table 1.

### 4.1 Verification of the Model

In order to verify the accuracy of the present numerical model, the modified case of the simulated results is then validated against the numerical results with the same geometric model obtained by He, Minoru, Ryu, Himeno, and Kawamura (2004). The axially symmetrical case of two-layer skin tissue with a cancerous tumor is used in the validation case. In the validation case, the skin is exposed to laser irradiation with an intensity of 1.4 W/mm$^2$. The results of the selected test case are illustrated in Fig. 3 for the temperature distribution in the skin with a cancerous tumor. Good agreement of the temperature distribution with elapsed times between the present solution and that of He, Minoru, Ryu, Himeno, and Kawamura (2004) is clearly shown. This favorable comparison gives confidence in the accuracy of the present numerical model. It is important to note that there may be some errors occurring in the simulations that are generated by the input thermal properties, optical properties, and numerical scheme.

### 4.2 Temperature distribution

Figures 4 and 5 demonstrate the temperature changes of the irradiation center with elapsed time. It can be seen for all graphs that the temperature rapidly increase in the early stage and gradually rise up afterwards and finally approach a steady-state value. The heating speed differs with different wavelengths as well as laser intensities. The laser with a wavelength of 532 nm and laser intensity of 2 W/mm$^2$ corresponds to the highest speed of heating up.

Figure 4 shows the skin temperature changes (at the position of z=0, r=0) with elapsed
time at the different wavelengths when the skin is irradiated by a laser with intensities of (a) 1, (b) 1.5, and (c) 2 W/mm$^2$, respectively. The laser with a wavelength of 532 nm produces the highest temperature value for all intensity levels while the maximum temperatures for the wavelengths of 755 nm and 800 nm do not different. This is because at the wavelength of 532 nm, the skin has a greater absorption coefficient than that of 755 nm and 800. When the skin is exposed to the laser, the absorbed energy is converted into thermal energy, leading to temperature increases. Moreover, it can be seen that greater intensity provides a faster temperature increase as well as faster heating up. Figure 8 shows the result of the skin temperature changes (at the position of $z=0, r=0$) with elapsed time at the different laser intensities when the skin is irradiated by a laser with wave lengths of 532, 755, and 800 nm, respectively. The laser with a wavelength of 532 nm has the highest maximum temperature for all intensity levels. Higher intensities provide faster temperature increases as well as faster heating up, similarly to the results illustrated in Fig. 4.

Figures 6 to 8 indicate the temperature fields in skin tissues after the laser-induced thermotherapy process with three different wavelengths (532, 755, and 800 nm) and with three different intensities (1, 1.5, and 2 W/mm$^2$) for three different durations of irradiation: 30, 60, and 600s. Overall, these simulated results show that the temperature increase is highest at the center of the irradiation region and decreases from the irradiation region towards the surrounding healthy tissues for all wavelengths and laser intensities after the application of laser irradiation. The effect arises from the fact that heat from the hot spot zone in the central region will diffuse to the cold region in the surrounding tissues, and in particular, the longitudinal temperature distribution affects the heat diffusion through the different layers of tissues. It is remarkable that the temperature decreases with increases in the wavelength and
that the temperature in the center of the irradiation region decreases with increasing wavelength. There are two steps during laser-induced thermotherapy: first, the tissue is heated directly within the optical absorption depth, and second, heat diffusion to the deeper tissue occurs. Generally, in the early stage of laser-induced thermotherapy, the heat diffusion does not reach very deep into the tissues. After that, heat diffusion has sufficient time to spread deeper into the tissues. The temperature fields in skin tissues under each test condition differently occur due to the effects of optical and thermal properties and absorption coefficient of the epidermal layer, which varies with wavelength. In addition, the blood perfusion rate and the skin surface cooling also affect the temperature fields in skin as well. These parameters play an important role in releasing the excess heat in the skin.

Figure 6 represents the temperature field at the early stage of the laser-induced thermotherapy process (30s). The skin depth heating effect causes a major part of the laser irradiation to be absorbed within the leading edge of the layered tissue for all wavelengths and laser intensities. Owning to the laser energy absorbed, the temperature distribution within the tissue decays slowly along the propagation direction, following Beer-Lambert's law, as described by Eq. (6). It is noted that in the case of the short wavelength (532 nm), it has the highest temperature value and has wider hot spot area at the skin surface for all intensity levels. This is because the epidermal layer has a higher value of the absorption coefficient than the dermal and subcutaneous fat layers. Therefore, the outer surface, where the epidermal layer is located, can absorb more energy at this wavelength than at other wavelengths because of the high value of its absorption coefficient.

Figure 7 presents the temperature distribution in the skin after 60 s of laser irradiation. The wavelength of 532 nm produces the highest temperature at the skin surface due to its
highest absorption coefficient value. In contrast, the wavelengths of 755 and 800 nm provide
more penetration depth due to the higher thermal conductivity in the dermal layer, as evidenced
by the intensity of 1W/mm$^2$ seen in Fig.7 (a). Meanwhile the increase of the laser intensity
level leads to more energy absorption and causes an increase in the temperature at the skin
surface because the skin surface is directly irradiated by the laser. In this case, there is not
enough time for heat diffusion to take place. So the position of the maximum temperature
changes to the skin surface, as shown in Figs.7 (a) and (b).

However, as time progresses (Fig. 8), the wavelength of 532 nm still has the highest
temperature value for all three intensities when compared with the wavelengths of 755 and 800
nm. Moreover, the hot spot area of the wavelength of 532 nm is wider and has a higher
temperature at the skin surface as a result of its greater absorption coefficient value of outer
skin layer. After further irradiation time up to 600 s and increases in intensities up to 1.5 and 2
W/mm$^2$, the maximum temperature with wavelengths of 755 and 800 nm still occurred deep
inside the skin layer, as illustrated in Figs.8 (b) and (c), in contrast with the case of the short
wavelength, where the upper surface or skin surface always has a higher temperature than the
tissue deep inside the skin. This is because the effect of thermal conductivity plays an
important role in the conductance of the laser energy absorbed and the irradiation time is long
enough for diffusion of heat into the deeper layer. This phenomenon has not been discussed in
detail in previous works. In this study, wavelengths of 532, 755, and 800 nm and laser
irradiation intensities of 1, 1.5, and 2 W/mm$^2$ on skin type I–II for an irradiation time of 600 s
served as an affiant in enhancing thermal response in laser-induced thermotherapy throughout
the treatment process without side effects on the healthy tissue in the surrounding area.

Regarding the 2D plot of temperature distribution, which is based on a 2D
axisymmetrical model of the layered skin, as shown in Fig. 7, the simulated temperature changes in layered skin in more planes, especially the hot spot zone in the layered skin under the same conditions, can again be recast in a 3D plane with also under the assumption of a 2D axisymmetrical plane, as shown in Fig. 9. In the figures, it can be clearly observed that the skin surface is sensitive to the thermal response during the laser-induced thermotherapy process.

5. Conclusion

Numerical simulation of the transient temperature distribution within the layered skin tissue during laser-induced thermotherapy under different treatment conditions has been performed. It was found that the temperature distribution and laser penetration depth depend strongly on the wavelength, laser irradiation intensity, and skin color type. Under the same intensity and irradiation time, an increase in the laser wavelength leads to decreases in the skin temperature. It is found that the absorption of laser energy is significantly higher in the shorter wavelength range (532 nm). The level of laser intensity affects the temperature increase within the tissues, with the higher intensity level leading to a higher temperature within the tissues.

Further, the darker skin color (skin type V–VI) absorbs more laser energy. However, laser absorption by melanin in the skin decreases with increasing wavelength. The absorption coefficient of white skin (skin type I–II) is lower than that of dark skin (skin type V–VI), and therefore the dark skin color (skin type V–VI) can absorb more energy than white skin. Consequently, epidermal damage occurs more easily in dark skin types compared to white skin types. As a result, special care should be taken when treating patients with dark skin (skin type V–VI), and the intensity level should be reduced in order to reduce unwanted thermal injury. Thus, as appropriate settings are necessary for highly effective and safe treatment, the proper laser for dark skin types is a longer wavelength laser because of its lower energy absorption.
characteristic. In addition, it is found that higher laser intensity results in higher absorption of laser energy by the skin. Moreover, this study also shows that the irradiation time has a marked influence on the temperature increase in the skin tissue.

The obtained results contribute to the understanding of the realistic situation and prediction of the temperature distribution in skin tissue during laser-induced thermotherapy under different treatment conditions.

In this study, it is concluded that the parameters could become significant factor to the transport phenomena in the human skin tissues when exposed to laser irradiation. However, for all of cases, only bioheat theory with optical and thermal properties are performed. Thus, more complicated theory should be taken for more accuracy of the simulation results.

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Table 1 Thermal properties and optical properties of tissues (Bhowmik et al., 2015; Chen et al., 2014; Aguilar et al., 2002; Tseng et al., 2009)

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Fig. 1 The skin model: (a) 3D skin model with laser irradiation, (b) 2D skin model with laser irradiation.

Fig. 2 Boundary conditions and physical domain.
Fig. 3 Comparison of the calculated temperature distribution with the temperature distribution obtained by He et al. (2004).

Fig. 4 Temperature changes (at the position of z=0, r=0) with elapsed time at different wavelengths (532, 755, and 800 nm) for skin color type I–II under laser irradiation with intensities of (a) 1, (b) 1.5, and (c) 2 W/mm$^2$, respectively.
Fig. 5  Temperature changes (at the position of z=0, r=0) with elapsed time at different laser intensities (1, 1.5, and 2 W/mm$^2$) on skin color type I–II under laser irradiation with wavelengths of (a) 532, (b) 755, and (c) 800 nm, respectively.
Fig. 6  Temperature distribution in the skin after laser irradiation with three different wavelengths of 532, 755, and 800 nm, where the laser intensities are set to (a) 1, (b) 1.5, and (c) 2 W/mm$^2$ (at t = 30 s).
Fig. 7  Temperature distribution in the skin after laser irradiation with three different wavelengths of 532, 755, and 800 nm, where the laser intensities are set to (a) 1, (b) 1.5, and (c) 2 W/mm² (at t = 60 s).
Fig. 8  Temperature distribution in the skin after laser irradiation with three different wavelengths of 532, 755, and 800 nm, where the laser intensities are set to (a) 1, (b) 1.5, and (c) 2 W/mm$^2$ (at $t = 600$ s).
Fig. 9  3D temperature distribution in the skin after laser irradiation with three different wavelengths of 532, 755, and 800 nm, where the laser intensities are set to (a) 1, (b) 1.5, and (c) 2 W/mm² at t = 60 s.