ER:YAG LASER TISSUE ABLATION AS A SURFACE VAPORIZATION PROCESS: A COMPUTATINAL MODELLING STUDY OF PULSE DURATION IMPORTANCE

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Abstract - Laser tissue ablation is taking important roles in surgical and medical applications. Therefore, it attained great attention from researchers to continuously enhance its therapeutic outcome. In this paper, Er:YAG laser tissue ablation was investigated using computational model. The model assumed laser tissue ablation a water surface vaporization process. Pulse duration was varied by two order of magnitudes and was found to have significant influence on the ablation outcome. The model predicted the existence of optimum pulse duration value that maximizes ablation rate while minimizes thermal damage. Determination of the exact value of such optimum will be investigated in future works.

Keywords - Surface Vaporization, Laser Model, Ablation, Thermal Damage, Pulse Duration

I. INTRODUCTION

Infrared (IR) and mid-infrared (mid-IR) lasers have been widely used in medical applications due to their ability to produce accurate and selective ablation [1, 2]. Water absorption peaks in the mid-infrared (mid-IR) wavelength region; thus mid-IR lasers have acquired special interest in the field of laser surgeries [3, 4]. The strong absorption coefficient of water in this region allows mid-IR laser to induce rapid clean ablative wounds with relatively small thermal damage in comparison with other laser modalities [5]. Mid-IR lasers are extensively employed in dermatology, specifically in aesthetic applications [6, 7]. Recently, lasers have been utilized to facilitate the delivery of topically applied drugs into subcutaneous layers [8]. However, the coagulated zone induced by lasers may add extra resistivity to drug delivery through tissues [9, 10]; hence, minimizing coagulated zone while maximizing ablation amount is, thus, a continuously sought goal.

In order to achieve such goal, a deep understanding of laser-tissue interactions and roles of each parameter involved in such interactions is needed. Despite the enormous investigations and advances in this matter, a complete clear understanding is still lacking. This is mainly due to the complexity of the thermo-mechanical mechanisms that are involved in the laser tissue ablation process [3]. Such mechanisms include surface vaporization [11], normal boiling [12], confined boiling [13], explosive boiling, photo-chemical decomposition [14], shock waves and recoil stresses [15]. In addition, one or more of these mechanisms can be simultaneously at work depending on the laser wavelength, pulse duration, and power. Tissues' composition and thermo-mechanical properties may also dictate which tissue ablation mechanisms shall be involved.

Experimental investigations of laser-tissue interactions are complex, expensive, and limited by the availability of tunable laser systems [16]. Therefore, theoretical models can be used to further the understanding of laser tissue ablation process by separately investigating the different parameters and mechanisms that are involved in this process. In this work, we are particularly interested in Er:YAG laser tissue ablation. Because Er:YAG laser is strongly absorbed by water, its interaction with water-rich biological tissues may resemble laser metal interactions [17]. Thus, Er:YAG laser tissue ablation is simulated here as a water surface vaporization process. A computational model is then developed to study the influence of pulse duration on ablation outcome.

II. MODEL DEVELOPMENT

More than 70 % of biological tissue consists of water [18]. Therefore, Er:YAG laser tissue ablation may be approximated as a process in aqueous medium. Surface vaporization depends on the surface temperature and is governed by the kinetic theory which, at equilibrium, can be defined as [19, 20]

$$J = \frac{\alpha\left(P - P_{amb}\right)}{\sqrt{2\pi m k_B T}} \qquad (1)$$

where J is the molecular flux at the surface, α is the vaporization coefficient, T is the surface temperature (in Kelvin), P is the vapor pressure (saturation pressure) at the surface temperature (T), P_{amb} is the partial pressure of water vapor in air (assumed 1 kPa), m is the mass of one molecule, k_B is the Boltzmann's constant. The velocity of tissue ablation is then equal to the surface recession velocity which is now xpressed as

$$v = \left.\frac{\partial x}{\partial t}\right|_{x=0} = \frac{JM}{\rho_w N_A} \qquad (2)$$

where M is the molar mass, ρ_w is the density of liquid water, N_A is Avogadro's number. Exploiting

Clausius–Clapeyron equation and substituting J from Eq. (1), Eq. (2) becomes

$$v = \frac{\alpha M}{\rho_w N_A \sqrt{2\pi m k_B T}} \left(P_b \exp\left(\frac{h_v}{R}\left[\frac{1}{T_b} - \frac{1}{T}\right]\right) - P_{amb} \right) \quad (3)$$

where P_b is the boiling temperature, h_v is the latent heat of vaporization per unit mass, R is the gas constant, T_b is the boiling temperature. Adopting a moving coordinate system that moves at the ablation velocity, the one-dimensional heat diffusion equation is written as

$$\rho_w C_p \frac{\partial T}{\partial t} = k \frac{\partial^2 T}{\partial x^2} + v \rho_w C_p \frac{\partial T}{\partial x} + \mu_a I_0 \exp(-\mu_a x) \quad (4)$$

where C_p is the specific heat, k is the thermal conductivity, μa is the absorption coefficient, I_0 is the laser irradiance at the surface. To solve Eq. (4), two boundary conditions and one initial condition are needed. These are

$$at\ t = t_0\ ,\ T = T_\infty = 20\ ^oC$$

$$at\ all\ t,\ at\ x = 0\ ,\ -k\frac{\partial T}{\partial x} = v\rho_w h_v \quad (5)$$

$$at\ all\ t,\ at\ x \to \infty,\ T \to T_\infty$$

Eq. (4) and (5) are discretized and solved using finite difference method, and their discretized forms are expressed as

$$T_i^{n+1} = \left(\frac{k\Delta t}{\rho_w C_p (\Delta x)^2} + v\frac{\Delta t}{\Delta x} \right) T_{i+1}^n + \left(\frac{k\Delta t}{\rho_w C_p (\Delta x)^2} \right) T_{i-1}^n + ... \left(1 - v\frac{\Delta t}{\Delta x} - 2\frac{k\Delta t}{\rho_w C_p (\Delta x)^2} \right) T_i^n + \mu_a I_0 \exp(-\mu_a x_i) \quad (6a)$$

$$T_1^{n+1} = T_2^{n+1} + v\frac{\rho_w h_v}{k} \quad (6b)$$

$$T_N^{n+1} = T_\infty$$

where i and n are indices for spatial position and time, respectively. Δx is the spatial step size, Δt is the time step length, and N is the index of the last node in the discretized domain. Δx and Δt values of 0.01 μm and 0.1 ns, respectively, were used to guarantee stability of the implemented numerical scheme. Thermal damage is computed using Arrhenius model as implemented for skin tissue in Ref. [13]. Thermo-physical properties of water were used [21, 22].

III. RESULTS AND DISCUSSION

Temperature profiles at the end of a single laser at three different selected values of pulse durations are shown in Fig.1. A subsurface temperature peak exists with all tested pulse duration values, with the largest peak corresponding to the shortest pulse duration. The maximum attained temperature is very close to the critical temperature of water (i.e. 373 ^oC). This may indicate that phase explosion could be initiated if irradiating with a shorter pulse or higher fluence [23]. The existence of such a peak is not surprising and is a direct result of the evaporative cooling effect at the surface [24].

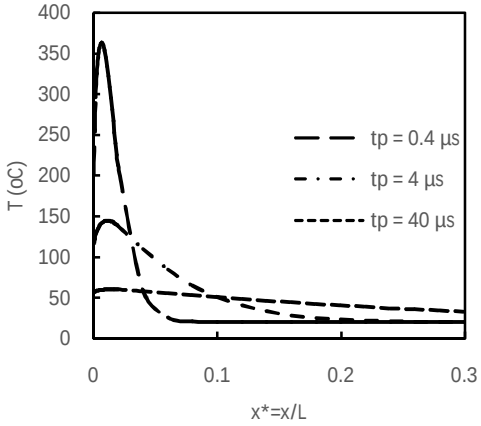
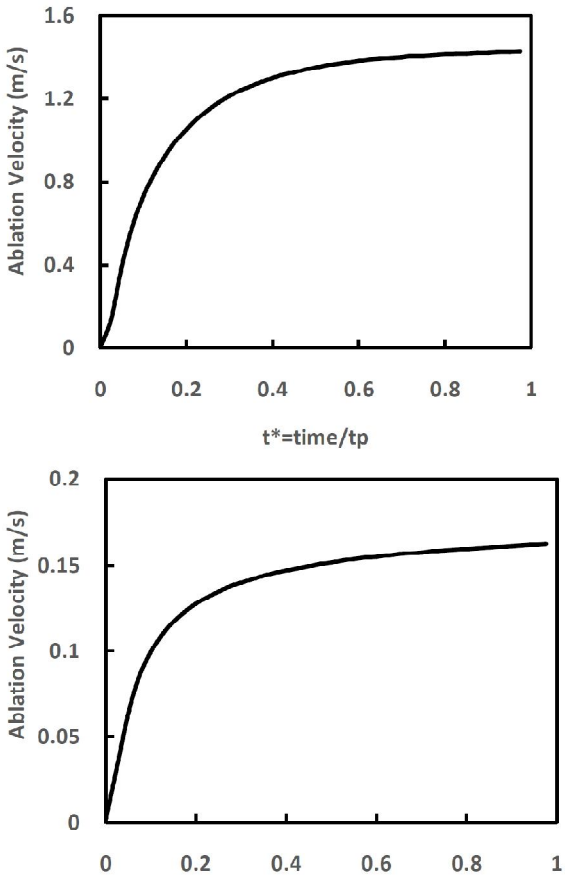


Fig.1 Predicted temperature profile at the end of single laser pulse expressed with respect to a dimensionless position variable x* defined as the ratio of position (x) to the length of the domain (L). Laser pulse fluence was 0.2 J/cm^2.

Ablation velocity increased exponentially during the laser pulse and saturated to a maximum value towards the end of the pulse as shown in Fig. 2. This maximum velocity is inversely proportional with the pulse duration where an order of magnitude increase in the pulse duration causes an order of magnitude decrease in the maximum velocity. The longer the laser pulse is, the more heat is lost, and the surface temperatures are, which lessens the vaporization rate and thus the ablation velocity.



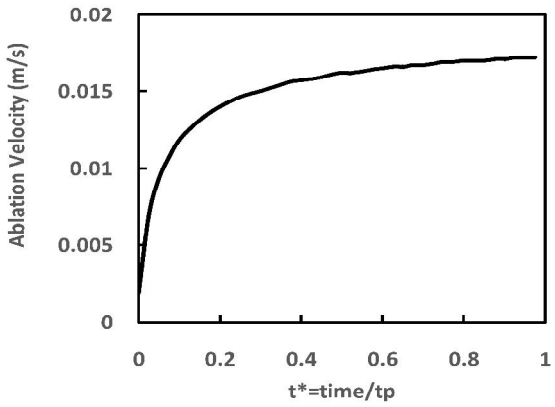


Fig.25 Ablation velocity during single laser pulse at three different selected values of pulse durations (t_p) (a) t_p =0.4 µs, (b) t_p =4 µs, and (c) t_p =40 µs. t* is the dimensionless time defined as the ratio of time to the pulse duration (tp).

However, such increase in ablation velocity doesn't necessarily guarantee proportional effect on the amount of tissue ablated. Fig. 3 shows that the surface position at the end of the longest laser pulse of 40 µs is more than 25 % deeper than the surface position at the shortest laser pulse of 0.4 µs. This implies that higher ablation efficiency is achieved with longer laser pulses. On the other hand, the length of propagated thermal damage at 4 µs laser pulse is larger than those at the shorter and longer laser pulses of 0.4 µs and 40 µs, respectively, as shown in Table 1. This suggests, in contrary to the thermal confinement threshold concept [3, 25], an existence of an optimum value of laser pulse duration that maximizes ablation rates while minimizes collateral thermal damages. Such value is expected to vary with laser wavelength and tissue properties. Determining the optimum pulse duration values is beyond the scope of the current study, but will be investigated in expanded future works.

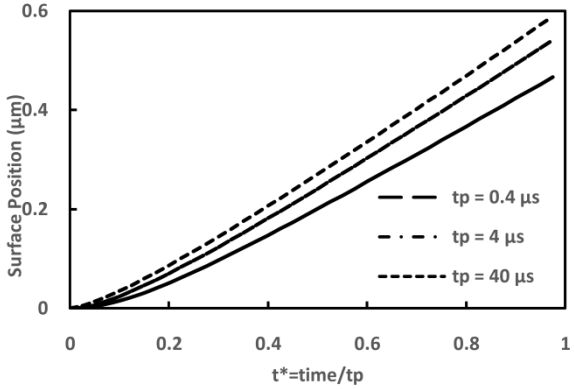


Fig.3 The position to which the surface has recessed to during single laser pulse

Table 1 Thermal damage depth at end of laser treatment

Pulse Duration	Depth of Thermal Damage
0.4 µs	0.4 µm
4 µs	1 µm
40 µs	zero

CONCLUSIONS

In this work, Er:YAG laser tissue ablation was modeled as a water surface vaporization process. Numerical solution was obtained using the finite difference method. Laser pulse durations were varied over a two order of magnitude range and were found to have significant effect on both ablation rate and thermal damage size. The simulation predicted an existence of optimum laser pulse duration that can maximize ablation rate and minimize thermal damage.

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