Published in final edited form as:

Comput Biol Med. 2013 December ; 43(12): 2278–2286. doi:10.1016/j.combiomed.2013.09.005.

Maxwell's Equations-based Dynamic Laser-Tissue Interaction Model

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Abstract

Since its invention in the early 1960's, the laser has been used as a tool for surgical, therapeutic, and diagnostic purposes. To achieve maximum effectiveness with the greatest margin of safety it is important to understand the mechanisms of light propagation through tissue and how that light affects living cells. Lasers with novel output characteristics for medical and military applications are too often implemented prior to proper evaluation with respect to tissue optical properties and human safety. Therefore, advances in computational models that describe light propagation and the cellular responses to laser exposure, without the use of animal models, are of considerable interest. Here, a physics-based laser-tissue interaction model was developed to predict the dynamic changes in the spatial and temporal temperature rise during laser exposure to biological tissues. Unlike conventional models, the new approach is grounded on rigorous electromagnetic theory that accounts for wave interference, polarization, and nonlinearity in propagation using a Maxwell's equation-based technique.

Keywords

finite element; laser-tissue interaction; photoacoustic; laser damage

Introduction

Theoretical modeling of the interaction between electromagnetic radiation and biological tissue has been extensively reviewed. It is well established that one of the primary mechanisms of interaction between laser radiation and biological tissue is thermal. The term “thermal interaction” represents a large group of interaction types, where the increase in

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local temperature is the primary parameter change. The tissue thermal response can be induced by either continuous wave (CW) or pulsed laser radiation. Depending on the exposure duration and the laser peak intensity, different effects such as coagulation, vaporization, carbonization, melting, and mechanical ruptures may be observed [1–10].

Biological tissues are turbid media and are characterized by their high scattering and absorption from representative chromophores: oxyhemoglobin, deoxyhemoglobin, melanin, collagen, and water. There are two complex quantities that are often used to describe optical properties: the complex index of refraction and the complex relative permittivity [11]. These two quantities are not independent and both are temperature dependent [12]. Changes in the optical properties of water play a dominant role in driving the distribution of light within tissue and have implications on photocoagulation and photoablation of tissues [13–15]. The present work was motivated by the need to develop a physics-based laser-tissue interaction model that accurately defines the internal electromagnetic fields within the exposed medium, taking into account that both absorption and scattering are dynamic and codependent on each other.

Lasers frequently used for dermatologic laser surgery are in the infrared (IR) region of the spectrum, where water is the targeted tissue chromophore. Because of its strong water absorption and surface deposition, the Er:YAG (2940 nm) laser is commonly used in surgery and cosmetic skin resurfacing. For skin resurfacing, typical specifications of the Er:YAG laser are 0.25ms pulse duration, 200–600 mJ pulse energy, and 3–9 mm spot size. Excimer lasers at ultraviolet (UV) wavelengths, which are well-absorbed by collagen, are being used in vision correction surgeries such as laser-assisted in-situ keratomileusis (LASIK) and photorefractive keratectomy (PRK). A typical example of an Excimer laser used in LASIK surgery emits 193 nm wavelength, and works at approximately 20 ns pulse duration, 100–250 mJ/cm² radiant exposure, with a beam 100 microns in diameter [16]. The dynamic changes in the optical properties of water at 193 nm (ArF excimer) and 2940 nm (Er:YAG) are well established [17, 18, 19].

The variations of tissue absorption and scattering with wavelength in the electromagnetic spectrum have been identified: absorption dominant, scattering dominant and a hybrid. The simplest case is the absorption dominant in the ultraviolet region and at wavelengths greater than 2700 nm. Beer's law has been used extensively to model the interaction of light with tissue in this region. The other extreme, where scattering dominates, applies to most tissues in the wavelength range 650–1300 nm. The radiative transfer theory-based Monte Carlo technique has been used to model the interaction of light with tissue in this region but with limitations [20, 21]. In the third interval, where absorption and scattering are both significant, it is difficult to make any general comments regarding the intensity distribution of laser radiation. This case applies to two spectral regions, one at roughly 300–650 nm and the other at 1300–2700 nm [20, 21]. Although the Monte Carlo technique has been widely used to model this region, the validity of the radiative transfer theory-based Monte Carlo method is yet to be determined. We conclude that accurately modeling laser-tissue interaction in all three regions will require the use of Maxwell's equations.

The Monte Carlo simulation of light scattering by a cluster of dielectric spheres is the most commonly used approach in modeling laser propagation through tissue. The radiative transfer theory-based Monte Carlo technique assumes independent scattering events between closely packed small dielectric spheres. This assumption modifies the physical nature of the original problem and fails to account for some of the essential elements of the physics in the problem, such as polarization and coherent interference. Therefore, the validity and scope of the Monte Carlo approach to treat the scattering of electromagnetic radiation by tissue is presently indefinite. The Monte Carlo technique was shown to deviate from the numerical solution of Maxwell's equations in computing the total scattering cross section [22, 23]. To maintain the wave aspect of light and take into account the effect of polarization and interference, the Maxwell's equations-based technique is the most credible and accurate approach to model the interaction of a laser with biological tissues.

Theoretical Model

Tissue optical properties are usually described in terms of scattering and absorption coefficients. In most models, the empirical average optical properties of small volumes of tissue are frequently used to simulate laser propagation in larger heterogeneous tissue volumes. In nearly all of the models, scattering and absorption are treated independent of each other, and as temperature independent static processes. Modeling the dielectric properties of biological tissue and changes to the spatial and the temporal profiles of those properties is essential for an accurate description of the more complicated laser-tissue interaction process. The optical properties of materials are obtained from their measured complex relative permittivity, $\varepsilon(\omega)$, expressed as:

$$\varepsilon(\omega) = \varepsilon' + i\varepsilon'', \quad (1)$$

where ε' is the real part of the relative permittivity of the material and ε'' is the imaginary part, known as the loss factor. Assuming that the material is nonmagnetic, the relationships between the complex relative permittivity ($\varepsilon(\omega) = \varepsilon' + i\varepsilon''$) and the complex refractive index ($n = n' + i n''$) are given by the following equations [11, 12]:

$$\begin{aligned} \varepsilon' &= n'^2 - n''^2 \\ \varepsilon'' &= 2 n' n''. \end{aligned} \quad (2)$$

Assuming small changes induced in the linear and the nonlinear index of refraction, Maxwell's equations lead to the following form of the Nonlinear Schrödinger Equation (NLSE) [24, 25]:

$$2 i k_0 \frac{\partial \vec{E}(x, y, z)}{\partial z} + \Delta_{\perp} \vec{E}(x, y, z) + \frac{(\omega_0)^2}{c^2} (n(x, y, z)^2 - n_b(x, y, z)^2) \vec{E}(x, y, z) + 2 \frac{(\omega_0)^2}{c^2} n_b n_2 |E|^2 \vec{E}(x, y, z) = 0, \quad (3)$$

where $\vec{E}(x, y, z)$ is the complex slowly-varying envelope of the electric field, k_0 is the wave number of the plane wave propagating in the background, $n_b(x, y, z)$ is the background refractive index, $n(x, y, z)$ is the linear refractive index which fluctuates as a function of temperature, and n_2 is the Kerr coefficient. The background refractive index is taken to be

the minimum value of the linear refractive index which has a temperature dependent profile that can be defined as [24]:

$$n(x, y, z) = n_o + \frac{\partial n}{\partial T} \Delta T. \quad (4)$$

Since $\epsilon(\omega)$ was assumed real, the linear refractive index in Eq. 3 is defined as

$$n(x, y, z) = \sqrt{\epsilon'(x, y, z)} \text{ and the relative index change should be very small } (n - n_b)/n_b \ll 1.$$

The form of the NLSE shown in Eq. 3 assumes real index of refraction and no significant absorption. This formulation works well when modeling laser propagation through a medium with no significant absorption and where scattering is the dominant process. But when modeling biological tissues such as the skin or the cornea, where a key optical absorber is water and significant scattering is also expected, the intrinsic optical properties of the tissue defined in Eq. 2 must be considered [26]. To account for absorption, scattering, and the temperature induced effect on both absorption and scattering, the general case for a dissipative medium must be considered by allowing the dielectric tensor to be a complex quantity [12, 24]. To account for the dissipative nature of biological tissue, the Maxwell's Equations-based paraxial slowly varying envelope approximation (SVEA) propagation of field envelope in a weakly-guiding optical wave guide with Kerr nonlinearity is considered:

$$2ik_0 \frac{\partial}{\partial z} \vec{E}(x, y, z) + \Delta_{\perp} \vec{E}(x, y, z) + \frac{\omega_0^2}{c^2} (\epsilon^{(1)} \cdot \vec{E}(x, y, z)) - k_0 \frac{(\omega_0)^2}{c^2} \vec{E}(x, y) + 2k_0 \frac{\omega_0}{c} n_2 |\vec{E}|^2 \vec{E}(x, y, z) = 0, \quad (5)$$

where $\vec{E}(x, y, z)$ is the complex slowly-varying envelope of the electric field, $k_0 = \omega_0 n(\omega_0)/c$ is the wave number of the plane wave $\approx e(i k_0 z - i \omega_0 t)$ propagating in the z direction, $\epsilon(\omega)^{(1)}$ is the complex dielectric tensor which fluctuates as a function of temperature, and n_2 is the Kerr coefficient which defines the nonlinear contribution, $n_{NL} = n_2 I$, to the refractive index. The intensity distribution of the laser beam is given by the square of the normalized optical field:

$$I(x, y, z) = |\vec{E}(x, y, z)|^2. \quad (6)$$

When the laser beam is turned on, it begins heating the tissue. The heating process is governed by the heat diffusion equation. The temperature distribution can be determined by solving the bio-heat diffusion equation with laser-induced heat sources [27].

$$\frac{\partial(\rho c T)}{\partial t} = \nabla \cdot (k \nabla T) + S - \rho_b c_b \omega_b (T - T_b), \quad (7)$$

where $S(x, y, z)$ is the source function that represents the power absorbed from the laser beam by a unit volume of the tissue at the point (x, y, z) . The temperature-dependent localized heat source $S(x, y, z, t)$ is defined as the multiplication of the temporal absorption coefficient $\mu_a(x, y, z, t)$ and the laser intensity $I(x, y, z)$ at that location.

$$S(x, y, z, t) = \mu_a(x, y, z, t) I(x, y, z, t). \quad (8)$$

The optical properties of tissue in the IR region are highly influenced by the absorption coefficient of water. Many researchers documented the thermally induced water absorption in the UV region [18]. Since water is the major constituent of all tissues, accounting for thermally induced water absorption is vital for accurate laser tissue interaction modeling in the IR and the UV portions of the spectrum. The refractive index change is a function of temperature and density changes. Based on available data, linear regression of pure water absorption coefficients with temperature is a good approximation [19]. Assuming a linear

refractive index as a function of temperature, $\frac{\partial n''}{\partial T} = \text{const}$, the temporal profile of the absorption coefficient $\mu_a(x, y, z, t)$ is given by:

$$\mu_a(x, y, z, t) = \mu_o + \frac{\partial \mu_a}{\partial T} \Delta T, \quad (9)$$

where $\mu_o = (4 \pi n''/\lambda)$, and n'' is the imaginary refractive index given by Eq. 2. The laser propagation approach described by Eqs. 2–9 describing the dissipative nature of biological tissue and the temperature induced changes to its optical properties will be used to define the temperature-dependent heat sources that form the laser induced temperature profile. Rapid heating of tissue by pulsed laser radiation leads to the generation and propagation of thermoelastic stresses. The propagation of the induced tensile stress in biological tissue is described by the general photoacoustic equation [28]:

$$\left(\nabla^2 - \frac{1}{v_s^2} \frac{\partial^2}{\partial t^2} \right) P(x, y, z, t) = - \frac{\beta}{\kappa v_s^2} \frac{\partial^2 T(x, y, z, t)}{\partial t^2}, \quad (10)$$

where $P(x, y, z, t)$ is the acoustic pressure at location (x, y, z) and time t , κ is the isothermal compressibility, v_s is the speed of sound, β is the thermal coefficient of volume expansion, and T is the temperature rise. The left-hand side of this equation describes the wave propagation, whereas the right-hand side represents the source term. Since the source term is related to the second time derivative of the temperature rise, a time-invariant temperature rise does not produce pressure. However, as indicated by the source term in Eq. 10, the profile of the induced pressure rise is expected to take the shape of the rate of change of the temperature rise profile, not the temperature rise profile.

Skin Model

To demonstrate our model, a three-layered image-based skin model was developed. The skin model was employed to simulate experiments and examine the skin thermal lesion threshold due to non-ionizing laser radiation. The top layer, the skin surface, is a very thin layer and contains no melanin and no blood. Under the skin surface are the absorbing layers which include the epidermis and dermis. The bottom skin layer, the hypodermis, is modeled as an extension of the dermis.

The absorption coefficient of the skin is defined by uniform baseline skin absorption, and a non-uniform absorption due to melanin in the epidermis layer. The main absorbing chromophores in the epidermis layer are water and melanin. The current implementation uses a normal skin histological section image to define melanin distribution. Thus, the nonuniform spatial distribution of the absorption coefficient in the epidermis layer is defined. The uniform baseline absorption coefficient of the epidermis layer is based on available temperature dependent optical properties of liquid water.

Assuming bloodless dermis, an average uniform absorption coefficient of the dermis layer is considered. Based on measurements (Ruiping Huang, S. Jacques, unpublished data) at the Oregon Medical Laser Center, the baseline absorption coefficient of dermis as a function of wavelength, λ , is approximated by the following expression[29]:

$$\mu_{baseline}=0.244+85.3 \times e^{-(\lambda-154)/66.2}. \quad (11)$$

Several approximate expressions for the melanosome absorption coefficient were developed by researchers. In particular, the following approximate expression was developed by researchers at the Oregon Health and Science University based on threshold for explosive vaporization of melanosomes by various pulsed lasers (from Jacques and McAuliffe 1987 and Jacques et al. 1996), was considered [30].

$$\mu_{mel}=1.7 \times 10^{12} \times (\lambda^{-3.48}). \quad (12)$$

As shown in Fig. 1, the domain of the problem is divided into a number of two-dimensional triangular elements connected via their vertices. Optical properties of each element are assigned based on the contents of each element. Regions where non-uniform spatial distribution of melanosomes is being considered are denoted by a dark color (Fig. 2).

Cornea Model

The absorption coefficient of the cornea, μ_{cornea} , is defined by assuming an initial uniform collective absorption coefficient due to water, μ_{water} , and due to collagen, $\mu_{collagen}$. Based on available optical properties of water and collagen, the localized absorption coefficient of the cornea changes with temperature.

Finite Element Formulation

Consider the coupled set of differential equations, Eqs. 5, 7, 10, and their boundary and initial conditions shown below (Eq. 13) to be solved in the two-dimensional (2D) domain defined by the schematic diagram shown in Fig. 1.

$$\begin{aligned}
E(\infty, y) &= 0, \quad E(x, \infty) = 0, \quad T(x, y, 0) = T_o, \quad T(\infty, y, t) = T_o, \quad T(x, \infty, t) = T_o, \\
P(x, y, 0) &= 0, \quad P(\infty, y, t) = 0, \quad P(x, \infty, t) = 0, \\
f(x, y) &= f(x, 0) = y_o \\
(P(x, y, t))_{y=y_o} &= f(x, y, t)|_{y=y_o} = P_{atm} \\
(E(x, y))_{y=y_o} &= f(x, y)|_{y=y_o} = \sqrt{(\omega_o \mu_o / 2k_o)} E(x, y_o) \\
\left(k \frac{\partial T(x, y, t)}{\partial x} + k \frac{\partial T(x, y, t)}{\partial y} \right)_{y=y_o} &= h (T(x, y, t) - T_e)|_{y=y_o} \\
\left(\frac{\partial P(x, y, t)}{\partial x} + \frac{\partial P(x, y, t)}{\partial y} \right)_{y=y_o} &= (P(x, y, t) - P_{atm})|_{y=y_o}.
\end{aligned} \tag{13}$$

The domain of the boundary value problem defined above can be divided into elements connected via their vertices. For a 2D problem, the domain can be divided into a number of two-dimensional triangular elements. For each element in the discretized domain, the unknown solution $\phi^e(x, y)$ within each linear triangular element is approximated as $\phi^e(x, y) = a^e + b^e x + c^e y$ where a^e , b^e , and c^e are constants to be determined (Fig. 1 - (b)). The Finite Element (FE) solution of the boundary value problem shown above was accomplished using the Galerkin method. The Galerkin method is a weighted residual technique that seeks a solution by weighting the residual of the differential equation. Assume an approximate solution $\hat{\phi}$ of the differential $L\phi = f$ in a given domain Ω . Substituting $\hat{\phi}$ for ϕ results in a nonzero residual $L\hat{\phi} - f = r \neq 0$. The best approximation for $\hat{\phi}$ reduces the residual r to the least value at all points in Ω . The weighted residual method enforces the condition

$R_i \int_{\Omega} w_i r \, d\Omega = 0$, where R_i denote weighted residual integrals and w_i are chosen weighted functions. For accurate solutions, the weighted functions are usually selected to be the same as those used for expansion of the approximate solution [31]. In solving the boundary value problem shown above, an approximate linear expansion of the unknown electric field vector $E^e(\vec{r}, \omega)$, the unknown temperature $T^e(x, y, t)$, and the unknown pressure function $P^e(x, y, t)$ within each triangular element is considered (Eq. 14).

$$\begin{aligned}
E^e(\vec{r}, \omega) &= \begin{pmatrix} E_r^e(\vec{r}, \omega) \\ E_i^e(\vec{r}, \omega) \end{pmatrix} = \begin{pmatrix} \sum_{j=1}^3 U_j^e(\omega) N_j^e(x, y) \\ \sum_{j=1}^3 V_j^e(\omega) M_j^e(x, y) \end{pmatrix} \\
T^e(x, y, t) &= \sum_{j=1}^3 N_j^e(x, y) T_j^e(t), \quad P^e(x, y, t) = \sum_{j=1}^3 N_j^e(x, y) P_j^e(t),
\end{aligned} \tag{14}$$

where $N_j^e(x, y) = (a_j^e + b_j^e x + c_j^e y) / 2\Delta^e$ is the interpolation or expansion function, and Δ^e is the area of the triangular element. Applying the Galerkin principle to the set of differential equations listed above generates a set of matrix systems of equations for the real and imaginary fields, the temperature, and the induced pressure at the nodes of the triangular elements. The solution of those matrix systems provide the temporal electric field, temperature rise, and induced pressure rise at all points in the solution domain.

Results

Corneal epithelial damage thresholds reported by McCally from a single-pulse exposure to 1540 nm radiation from an Er:YAG fiber laser for exposure durations ranging from 0.025 s to 0.24 s are shown in Table 1 [32]. We used our model to simulate corneal response to damage thresholds exposures reported by McCally. The transient center temperature responses predicted by the model during the 0.24 s, 0.1 s, 0.045 s, and 0.025 s exposure durations listed above are compared in Fig. 3 (a). The model-predicted peak temperatures are shown in Table 1.

The experimentally measured corneal ED₅₀ values for a 2000 nm wavelength radiation and 1.17 mm spot diameter reported by Chen et al. for a range of exposure durations are shown in Table 2 [33]. We used our model to simulate Chen et al. experiments. The model predicted transient center temperature response for all exposure durations are compared in Fig. 3 (b). The model-predicted peak temperatures and lesion sizes are listed in Table 2.

Threshold measurements for the 1540 nm and 1314 nm lasers reported by Cain et al. are shown in Table 3 [34]. No threshold data was reported by Cain for the 1540 nm wavelength, 2.0 mm spot diameter, and 50 ns pulse duration, but a minimum visible lesion (MVL-ED₅₀) measurement of a 3.5 J/cm² was reported by Lukashev et al. for a 1540 nm wavelength, 2.5–3.5 mm spot diameter, and 100 ns pulse duration [35]. We used our model to simulate both Cain et al. and Lukashev et al. experiments. The peak temperatures predicted by the model, when simulating Cain et al. and Lukashev et al. experiments are shown in Table 3.

Damage threshold to pigmented skin resulting from Q-switched, 40-ns, 694-nm ruby laser irradiation was reported by Polla et al. at 0.8 J/cm² [36]. We used our model to simulate Polla's experiment for 1- and 2-mm diameters at different levels of radiant exposures. Results of the simulations are shown in Fig. 4.

The ArF excimer laser (193 nm wavelength, 13 ns) induced surface temperature in vitro porcine cornea was reported by Ishihara et al. for 40, 90, and 135 mJ/cm² radiant exposure levels [37]. We used our model to simulate the corneal response to ArF excimer laser (193 nm, 13 ns) at all three levels of radiant exposure. The model predicted transient center temperature response for all three levels of exposures and are compared in Fig. 5(a). The model predicted temperature profile for the 40 mJ/cm² exposure level is shown in Fig. 5(b).

Simulating the 2940 nm laser interaction with the cornea, taking into account the dynamic nature of the tissue optical properties, revealed a substantial difference in temperature rise compared to the static case shown in Fig. 6 (a). The induced acoustic pressure profiles for the static and the dynamic case are shown in Fig. 6 (b) and Fig. 6 (c) respectively.

Discussion

The focus of this work is to develop a physics-based laser-tissue interaction model to reproduce experimental results. As illustrated in ref. 22 and 23, simulating light propagation in turbid medium using the Monte Carlo technique yields different results compared Maxwell's equations. As will be shown in this section, results generated using our

Maxwell's equations-based model is in agreement with experimental data. However, radiant exposure at the ionization and plasma formation threshold level is outside the scope of this model. It is also to be noted that this model does not take into account multiphoton absorption and free-carriers absorption.

Until recently, the temperature at which a laser-exposed cell dies was not determined empirically. Using microthermography in an in vitro model, Denton et al. [38] determined the temperature profile of cells at the threshold of cell death, as indicated by post laser exposure fluorescence imaging. Although the size of damage across the cell monolayer varied depending upon laser irradiance, the peak temperature of the cells at the boundary between live and dead zones is 53 ± 2 °C. Surprisingly, the same value was determined for cells exposed for 0.1, 0.25, and 1.0 s. Figure 7 illustrates the distinction between the central and boundary peak temperatures. By definition, exposure from a threshold laser irradiance (E_{TH}) would generate a minimal lesion such that the maximum temperature in the center (T_X) is equal to the boundary threshold temperature (T_B). As the size of the lesion expands, due to increased absorption or laser irradiance (E_I), T_X increases above T_B because the boundary of cell death occurs at the periphery of the lesion. The T_B is thus a measure of the biologically significant threshold peak temperature [38].

With this premise, we wanted to compare our model results to the biologically relevant threshold peak temperature of Denton et al. Using the corneal epithelial damage thresholds reported by McCally for Er:YAG (1540) laser exposures, our model predicted nearly constant peak temperatures of about 53 °C (Table 1). As defined by McCally, the damage threshold is at the center of a narrow bracket at which a 10% difference in irradiance between an exposure that produced a minimal lesion and an exposure that did not produce a lesion was established. As indicated by McCally, his computed temperature rise resulting from the threshold exposure with duration 0.24 s failed to meet his anticipated approximately constant temperature to within the experimental uncertainty (Table 1). McCally's center temperature (on beam axis) calculations were based on a time-dependent Green function solution to the heat equation for the case in which a Gaussian profile laser beam incident on a semi-infinite slab is absorbed according to Beer-Lambert law. It is to be noted that beside the fact that it does not take into account polarization and interference effects, Beer's law is designed to account for total attenuation and it cannot separate the absorption from scattering. For the 1540 nm wavelength both absorption and scattering are significant and the Maxwell's equations-based approach is the suitable technique for accurate modeling of the laser radiation intensity within the exposed tissue. As expected, for all threshold exposures reported by McCally, the model-predicted peak temperature is almost constant and is at the threshold level for cellular damage (53 ± 2 °C) established by Denton et al.

Simulating the 2000 nm laser interaction with the cornea using threshold data reported by Chen et al., resulted in model-predicted temperatures at the threshold level for cellular damage (53 ± 2 °C) established by Denton et al. for small beam size, but slightly above that threshold level for larger spot sizes. As indicated by Chen, uncertainties in measurements included errors in spot diameter and laser power. Laser beam profiles measured by a beam profiler were elliptical rather than circular and the spot diameters were estimated by taking

the average of the major and the minor diameters. This resulted in 14% uncertainty in calculated radiant exposures [33]. The diameter of the threshold lesion sizes reported by Chen et al. ranged from 0.3 to 0.6 mm (0.4 ± 0.12 mm). The temperature profiles generated by the model were scrutinized against the damage threshold established by Denton et al., and estimates of lesion size were extracted. We have thus identified a modeled boundary of cell death using the expected 53 °C threshold temperature. As shown in Table 2, for all exposure durations the model-predicted lesion sizes fall within the range of the measured threshold lesion sizes reported by Chen et al.

Simulating the 1540 and 1314 nm lasers interaction with the skin using damage thresholds data reported by Cain et al., the model-predicted peak temperatures for long pulse durations is almost constant and it is at the temperature threshold level for thermal damage (53 ± 2 °C) established by Denton et al (Table 3). The only exception to this is the 68.97 °C model-predicted peak temperature for the 1540 nm wavelength, 20 J/cm² level of exposure, and a 0.7 mm spot diameter. The high temperature predicted by the model at the 20 J/cm² level of exposure is a clear indication of a possible attenuation of the laser energy before it reaches the skin. As reported by Cain, flashes of light and pop sounds were observed and those are signs of breakdown occurring at or near the surface of the skin. Thus, the 20 J/cm² pulse energy must have been attenuated and not all of its energy penetrated the skin. The model-predicted peak temperature when simulating Cain et al. and Lukashev et al. nanosecond exposures are 45.123 and 43.017 °C respectively. At this temperature level no thermal damage is expected. As reported by Cain, the histological examination revealed signs of mechanical damage concentrated on the pigmented epithelial cells and no damage was observed on the superficial epidermal layers from the nanosecond pulse duration. The low temperature rise our model predicted is a confirmation that thermal damage is not to be expected at this level of exposure. However, there is a higher likelihood of mechanical damage at the dermal- epidermal juncture due to its weakness.

Simulating the interaction of the Q-switched (694 nm) ruby laser with the skin using the pigmented skin damage thresholds data reported by Polla et al., our model-predicted peak temperature at the measured exposure level for epidermal lesions and cellular necrosis are 45 °C and 49 °C, respectively (Fig. 4). Since no thermal damage is expected at this temperature level, mechanical damage is likely to take place on the pigmented epithelial cells. As concluded by Polla, the basic mechanism of damage to pigmented skin resulting from Q-switched, 40-ns, 694-nm ruby laser irradiation is specific to energy absorption by melanosomes [36]. According to Polla, at 0.8 J/cm², epidermal lesions were more marked and at 1.2 J/cm², the epidermal cells of black female guinea pigs formed confluent zones of cellular necrosis up to the stratum granulosum, and lesions were also present 0.5 mm deep in follicles [36]. As indicated by Polla, melanosomes disruption may be an indication of shock wave and/or cavitation resulted from rapid thermal expansion [36].

The use of ArF excimer laser in laser eye surgeries such as PPK and LASIK has been thoroughly investigated. The ArF excimer laser (193 nm) is commonly used for vision corrective eye surgery. However, the mechanism of its interaction with the cornea is still being debated between surface vaporization and thermo-elastic stress ablation. The common approach being used to model corneal tissue interaction with a laser assumes a homogenous

and static medium and describes beam attenuation according to Beer's law. It is to be noted that a wide range of corneal absorption coefficients at 193 nm, between 2700 and 39000 cm^{-1} , have been reported [39, 40]. The possibility of a temperature dependent corneal absorption coefficient in the ultraviolet region has also been investigated. Many of the current laser tissue interaction models are grounded on the assumption that collagen is the primary chromophore in the cornea at 193 nm. However, available data suggests an increase in the absorption coefficient of water at 193 nm as a function of temperature. Experimental data reported by Staveteig and Walsh suggested a shift in the water absorption band located at 160 nm to longer wavelength [18]. Simulating the ArF excimer laser (193 nm wavelength, 13 ns) interaction with the cornea at 40, 90, and 135 mJ/cm^2 radiant exposure levels, taking into account the dynamic aspect of water optical properties, our model-predicted peak temperature is in agreement with the experimentally measured temperature reported by Ishihara et al. for 40 and 90 mJ/cm^2 (Fig. 3). For a radiant exposure of 135 mJ/cm^2 , our model-predicted peak temperature is a little higher compared to the experimentally measured peak temperature reported by Ishihara. This level of exposure, 135 mJ/cm^2 , is almost three times the ablation threshold, 40 mJ/cm^2 , and our current model does not take into account mass transfer. Therefore, energy loss due to ablation has not been taken into account and the model-predicted temperature is expected to be higher than the experimentally measured temperature.

To demonstrate a fundamental difference between our new dynamic model and the traditional static approach, we simulated the interaction of the 2940 nm laser with the cornea and examined the differences between the static and the dynamic approaches. Typically, a tissue optical properties measurement is taken at room temperature. The reduction in the IR absorption coefficient of tissue has been investigated by various researchers and several have concluded that the absorption peak of water at 2940 nm decreases and shifts towards shorter wavelengths. Results from a study done by Shori and coworkers for a wavelength of 2940 nm concluded that the absorption coefficient remains nearly identical to the room-temperature value of 13000 cm^{-1} until volumetric energy density of 0.02 kJ/cm^3 is reached and then decreases monotonically, reaching 2000 cm^{-1} at 20 kJ/cm^3 [41]. As shown in Fig. 4 (a), simulating the 2940 nm laser interaction with the cornea, taking into account the dynamic nature of the tissue optical properties, revealed a substantial difference in temperature rise compared to the static case. Another significant dissimilarity between our new dynamic modeling approach and the static approach is the difference in the induced acoustic pressure profile. As indicated by the source term in Eq. 10, the profile of the instantaneous rise in pressure is expected to take the shape of the rate of change of the temperature rise profile. Since the absorption coefficient decreases monotonically as a function of energy deposited in a given location, for a Gaussian beam the source term in the bio-heat equation decreases fastest at the beam center location. As we move away from the beam center the heat source term is expected to continue to decline but at a slower rate. As shown in Fig. 4 (b), the static approach leads to the formation of a Gaussian acoustic pressure profile resulting from a Gaussian beam profile exposure. However, changes to the absorption coefficient profile lead to successive formation of an induced acoustic pressure profile with its peak forming a circular ring around the beam center (Fig. 4 (c)). It is also

observed that the instantaneously induced thermoelectric pressure level continues its decrease as we move from the peak pressure location radially towards the beam center.

The physical basis for scattering and absorption of electromagnetic wave is related to the atomic and molecular composition of the medium being exposed. Under the effect of an electromagnetic wave, charges are set into oscillatory motion by the electric field of the incident wave and a dipole moment is induced in each region. The portion of the energy transferred by the accelerated charges into other forms of energy such as heat is characterized as absorption. The secondary electromagnetic energy radiated by the accelerated charges is known as the scattered radiation. At a given point in the medium, the total scattered field is the superposition of all scattered waves taking into account their phase differences. While the phases of the scattered waves depend on the direction, size, and particle shape, the phase and the amplitude of the induced dipole moment depends on the material [42]. The material response to the incident field is described by its permittivity, which is directly related to the electric susceptibility, the constant of proportionality that relates the incident field to the induced polarization. Thus, accurate characterization of the medium response to an electromagnetic field involves calculating the induced polarization at a given location in the medium taking into account the superposition of the scattered waves and their contribution to polarization. The Monte Carlo approach disregards polarization and it assumes independent scattering events by isolated particles. It also disregards the secondary electromagnetic energy radiated by the accelerated charges. Thus, the Monte Carlo approach assumes particles are excited by the incident field and it transforms the incident field into a scattered field.

Summary

We have presented a Maxwell's equations-based laser-tissue interaction model that accounts for wave interference, polarization, and nonlinearity in laser propagation. This new approach incorporates the physics required for accurate modeling of absorption and scattering of light as it propagates through tissue. Our model also takes into account the dynamic nature of tissue optical properties and their impact on the induced temperature and pressure profiles. Model predictions for the interaction of lasers at wavelengths of 193, 694, 1314, 1540, 2000, and 2940 nm with biological tissues were generated and comparisons were made with available experimental data for the cornea or the skin. Good agreement between model and experimental results established the validity of the model.

Acknowledgments

The authors acknowledge financial support from TASC Inc. In addition, we recognize Dr. Robert J. Thomas of the Air Force Research Laboratory (Fort Sam Houston, San Antonio, Texas) for his input. Partial funding of this work has also been provided by NIH/NIGMS MBRS-RISE GM60655 and NSF-PREM Grant No. DMR-0934218

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Fig. 1.

A schematic diagram of the (a) 2D FE domain in Cartesian coordinates system (b) linear triangular element.

Fig. 2.
A schematic diagram of the skin model.

Fig. 3. Maxwell model transient center temperature response for (a) 1540 nm wavelength for 0.24 s, 0.1 s, 0.045 s, and 0.025 s exposure durations, 1-mm spot diameter (b) 2000 nm wavelength for 4 s, 2 s, 1 s, 0.5 s, 0.25 s, and 0.1 s exposure duration, 1.17 mm spot diameters.

Fig. 4. Peak temperature as a function of radiant exposure and damage thresholds for 694-nm wavelength, 1- and 2-mm diameters, and 40-ns pulse duration.

(a)

(b)

Fig. 5. Maxwell model transient center temperature response for (a) 193 nm wavelength for, 13 ns exposure duration, 40 mJ/cm², 90 mJ/cm², and 135 mJ/cm² radiant exposures (b) Temperature profile at the end of a 193 nm wavelength, 40 mJ/cm² radiant exposure, and 13-ns pulse duration.

(a)

(b)

(c)

Fig. 6.

(a) Transient center temperature response for 2940 nm wavelength for 0.3 s exposure duration, and a 1.0 mm spot diameter. (b) Induced pressure profile for static simulation (c) Induced pressure profile for dynamic simulation

Fig. 7.
Illustration of threshold temperature at the center and boundary of cell death.

Table 1

Corneal damage thresholds and model predicted temperature rise for single-pulse exposure to 1540 nm radiation [32]

Exposure duration (s)	Radiant exposure (J/cm ²)	Irradiance (W/cm ²)	McCally's computed Center Temperature (°C)	Model prediction Center Temperature (°C)
0.24	13.8	57.4	64.4	53.03
0.1	9.43	94.3	59.4	54.01
0.045	6.75	150	54.5	53.72
0.025	4.60	184	49.4	50.71

Table 2

Model predicted temperatures at the center of the beam and lesion diameters for threshold irradiation of the cornea to 2000 nm wavelength [33]

Exposure Duration (s)	ED ₅₀ (mW)	Model predicted diameter of lesion (mm)	Model predicted Center Temperature (°C)
4	73.2 ± 0.8	0.22	57.9
2	87.6 ± 1.9	0.28	58
1	114.9 ± 5.1	0.32	61.25
0.5	116.5 ± 1.1	0.55	56.2
0.25	154.1 ± 3.6	0.49	55.94
0.1	236.1 ± 2.7	0.53	55.1

Table 3

Skin damage thresholds and model predicted temperature rise for single-pulse exposure of 1540 and 1314 nm radiation [34]

Skin Exposures Single pulse	MVL-ED ₅₀ (J/cm ²) 1540 nm 24 hrs	MVL-ED ₅₀ (J/cm ²) 1314 nm 24 hrs	Model Prediction Center Temperature (°C)	
			1540 nm	1314 nm
0.7-mm dia. Spot 350 μ s or 600 μ s	20 (21–18)	99 (112–86)	68.97	56.49
1.3-mm dia. Spot 350 μ s or 600 μ s	8.1 (8.7–7.5)	83 (81–85)	50.608	54.79
2.0-mm dia. Spot 31 ns or 50 ns	No Data	38.5 (51–31)	–	45.123