Turkish Journal of Electrical Engineering & Computer Sciences

http://journals.tubitak.gov.tr/elektrik/

Research Article

Turk J Elec Eng & Comp Sci
(2016) 24: 807 – 819

doi:10.3906/elk-1307-95

Study of temperature distribution in light–tissue interaction using the FEM

Ahmet Yasin ÇİTKAYA*, **Şaban Selim ŞEKER**
Department of Electrical and Electronics Engineering, Faculty of Engineering, Boğaziçi University, Bebek, İstanbul, Turkey

Received: 13.07.2013 • **Accepted/Published Online:** 20.10.2013 • **Final Version:** 23.03.2016

Abstract: In this paper, the numerical modeling of interaction between light sources and living biological tissue is studied. One of the most important results of physical interactions of light with biological tissue are thermal changes, which are a consequence of absorption of light. The severity of the effects is dependent upon several factors, including exposure duration, wavelength of the beam, beam energy, tissue type, and the area that is exposed to the beam. The temperature response of tissue irradiation is governed by the extensively used Pennes bioheat equation in a 3-dimensional space. The Pennes equation is solved initially for a single-layer human skin model, thanks to which temperature distribution can be predicted using different laser light radiation burns at steady state. Furthermore, a multilayer model is used to investigate the effects of the energy of the beam and exposure duration. The numerical solutions are obtained by the finite element method (FEM), in which the geometry studied is divided into a finite element mesh. This method gives a better description of the geometry with a smaller number of nodes when compared with the finite difference time domain (FDTD) method. Using the FDTD method, which may require spatial step ranges to achieve convergence, makes the computational simulation time increase dramatically. On the contrary, less memory space and disk space with shorter run times make the FEM more advantageous than other numerical solution techniques and that is why this method has been chosen for this study. Temperature contours and penetration depths are plotted in 3-dimensional spaces. In this work it is demonstrated that by adjusting the exposure duration, wavelength of the beam, energy of the beam, and the area and type of tissue, temperature effects of light on tissue can be altered.

Key words: Temperature distribution, finite element method, bioheat equation

1. Introduction

In certain medical treatment applications, laser light sources are used to generate optothermal effects in tissue. In these treatments, the most important issue is to control the temperature increase and distribution over the living tissue; otherwise, high temperatures could potentially cause undesired thermal damages in the surrounding pathology. The thermal response of the tissue that is exposed to laser light mainly depends on the optothermal properties of both the tissue and the light source, such as exposure duration, wavelength of the beam, energy of the beam, and the area and type of tissue exposed to the beam. In medical treatment applications with light, prediction of the thermal response of tissue can be realized by utilizing simulation tools such that selection of laser light dosage and exposure duration on the subject tissue can be done appropriately [1].

Temperature distribution in biological tissues can be analyzed via certain mathematical models, one of the most common of which is the Pennes bioheat equation [2]. The reason why the Pennes bioheat equation

*Correspondence: citkaya@boun.edu.tr

is still widely in use, despite many alternative improved models in the literature, lies behind its simplicity and accuracy provided in numerical analysis.

With the advancement of computational technology, mathematical modeling of temperature distribution in human tissue is become prevalent. In order to adopt the bioheat equation to investigate temperature distribution, several mathematical models are being used. Although several analytical and numerical methods have been developed in the literature, finite difference and finite element methods are the commonly used ones.

In this paper, the temperature distribution during laser treatment is predicted with a newly developed human skin model. The finite element method (FEM) is applied for modeling of numerical simulations to analyze the temperature changes in human skin. The objective of this research is modeling the laser–tissue interaction to optimize the effective parameters in order to understand optimal laser dosage to prevent damage. The severity of the damage depends on several factors, including exposure duration, wavelength of the beam, energy of the beam, and the area and type of tissue exposed to the beam.

The modeling of the physical interaction of light with biological tissue was attempted in some recent research papers. The objective consists mainly of two concerns: the optical propagation of light, and its thermal distribution in biological tissue. Different approaches, such as the seven-flux model [3], the Kubelka–Munk method [4], and the Monte Carlo method [5], have been developed for optical propagation, whereas the thermal modeling part has been given special attention with the usage of the most widely known approach: the Pennes bioheat equation. This extensively used bioheat equation has better resemblance to the experimental results than the effective thermal conductivity [6]. Simulations of biological tissue with the Pennes equation are quite good for microvessels; however, in the case of large vessels, a more complex method is needed [7].

Although the thermal problem in various biological tissue conditions has been worked on, there has been little attention paid to the numerical method required for the solution of the chosen equation. Studies in the literature show that the control volume-based finite difference procedure can also be used as an alternative for the FEM used to solve the bioheat transfer equation. The finite difference time domain (FDTD) method is selected as the main numerical method by some researchers for studying certain problems in tissue interfaces. Nevertheless, studies on thermooptical modeling use either the FEM or a simple, explicit one-step FDTD method. Convergence is achieved in the simple, explicit one-step FDTD method via limited time and spatial step ranges. However, the need to use very small steps for reaching a solution makes the computational simulation time rise drastically. Moreover, numerical predictions near the boundaries of the spatial numerical grid can be invalid due to the fictitious boundaries when the continuity condition is imposed. The FEM, on the other hand, enables the simulation of more complex geometries and usually results in relatively satisfactory outputs in terms of accuracy.

This study proposes the utilization of the FEM algorithm to improve the numerical solution of the bioheat equation. In the first step, the optothermal model, which employs the Beer–Lambert law for optical propagation and the Pennes equation for spatiotemporal temperature distribution, is explained. Afterwards, the numerical method proposed is presented, in which the simulation times are expected to be much shorter than with conventional FDTD techniques in spite of the improved characteristics implemented. To generate a representative instance, the mathematical modeling is finally applied to irradiation of the skin.

2. Mathematical modeling

The relationship between the absorption of light in a purely absorbing medium and the thickness of the medium was first determined in 1729 by Bouguer. In 1760, Lambert derived the following mathematical expression for

the relationship, known as the Lambert–Bouguer law. For incident intensity I_0, the transmitted intensity I through a distance l will be:

$$I = I_0 e^{-\mu_a l}. \tag{1}$$

The absorption coefficient μ_a can thus be interpreted as the probability that a photon will be absorbed by the medium per unit length. The reciprocal of the absorption coefficient, known as the absorption length, is the distance required for the intensity of the beam to fall to e^{-1} of the initial intensity.

2.1. Bioheat equation

Light absorbed by the tissue results in local temperature increase. The bioheat equation describes the tissue heat transfer due to the deposited light taking into account conduction, convection by blood, and possible heat sources (Q). The bioheat transfer equation was first introduced by Pennes in 1948 to model heat transfer in perfused tissue such that:

$$\delta_{ts}\rho C\frac{\delta T}{\delta t} + \nabla\cdot(-k\nabla T) = \rho_b C_b \omega_b (T_b - T) + Q_{met} + Q_{laser}, \tag{2}$$

where:

ρ = density of tissue (kg cm^{-3}),

C = specific heat of tissue (J kg^{-1} $^{\circ}\mathrm{C}^{-1}$),

T = temperature ($^{\circ}$C),

k = thermal conductivity of tissue (W cm^{-1} $^{\circ}\mathrm{C}^{-1}$),

C_b = specific heat of blood (J kg^{-1} $^{\circ}\mathrm{C}^{-1}$),

w_b = volumetric perfusion rate (kg s^{-1} cm^{-3}),

T_b = temperature of arterial blood ($^{\circ}$C).

The heat conduction term in the Pennes bioheat equation considering blood perfusion and metabolic heat generation is based on the classical Fourier law that implies an infinite thermal propagation speed [8]. Convection between tissue surface and surrounding air, radiation, and evaporation are processes that have to be modeled as appropriate boundary conditions.

$$q_{conv} = h_{amb}(T_s - T_{amb}), \tag{3}$$

where q_{conv} is the heat flux due to convection and T_s is the surface skin temperature. Assuming that the average temperature at the surface of the skin is 34 $^{\circ}$C, and using values of $h_{amb} = 10$ and $T_{amb} = 25$ $^{\circ}$C, q_{conv} is calculated to be 90 Wm^{-2} [9].

The laser energy radiated at each end of a fiber acts as a spatial heat source defined by:

$$Q_{laser} = aI_0 e^{-k_{dis}A}. \tag{4}$$

In the above equation, the variable I_0 is the intensity of the laser. I_0 is calculated by assuming an area of tissue and an output power of the laser. The variable k_{dis} is the dissipation constant and describes the loss of energy over time in the tissue. The variable A is the normal distance between the end point of the fiber and points in the laser beam.

3. FEM simulations for single-layer tissue model

The numerical solution relies on the FEM, in which the geometry studied is divided into a finite element mesh with computer software that makes it possible to numerically solve partial differential equations. Thus, instead of trying to solve a highly nonlinear problem on the entire geometry, an approximate solution is sought in each element. If this element is considerably small, the physical problem is assume3d to vary linearly.

The computational model is simulated for a 5 × 5 × 3 mm single layer of tissue. The bottom surface of the tissue is assumed to be located within the body core. We applied several laser light sources including CO_2, Nd:YAG, and ArF excimer lasers. For the first simulation, a 10,600 nm CO_2 laser in the IR region is used and an absorptivity value of 79,200 m^{-1} is used for skin. Additional simulation results are given for a 1064 nm Nd:YAG laser in the NIR region, again for skin, and a 1000 m^{-1} absorptivity coefficient is used. Finally, in the UV region, a 193 nm ArF excimer laser is used and an absorptivity value of 255,800 m^{-1} is accepted for the human eye lens. Penetration depths and temperature distributions obtained from the simulations [10] for the three types of laser sources are given in Figures 1–3.

4. Multilayer modeling of tissue

Optical properties of skin tissue hold a vital place not only for prediction of light exposure and interpretation of diagnostic data for medical and surgical applications [11], but also for new technology development in optothermal therapy. Though there are many studies on the assessment of optical parameters of biological tissue, they are not studied in the wide wavelength range [12]. Despite the abundance of scientific studies on

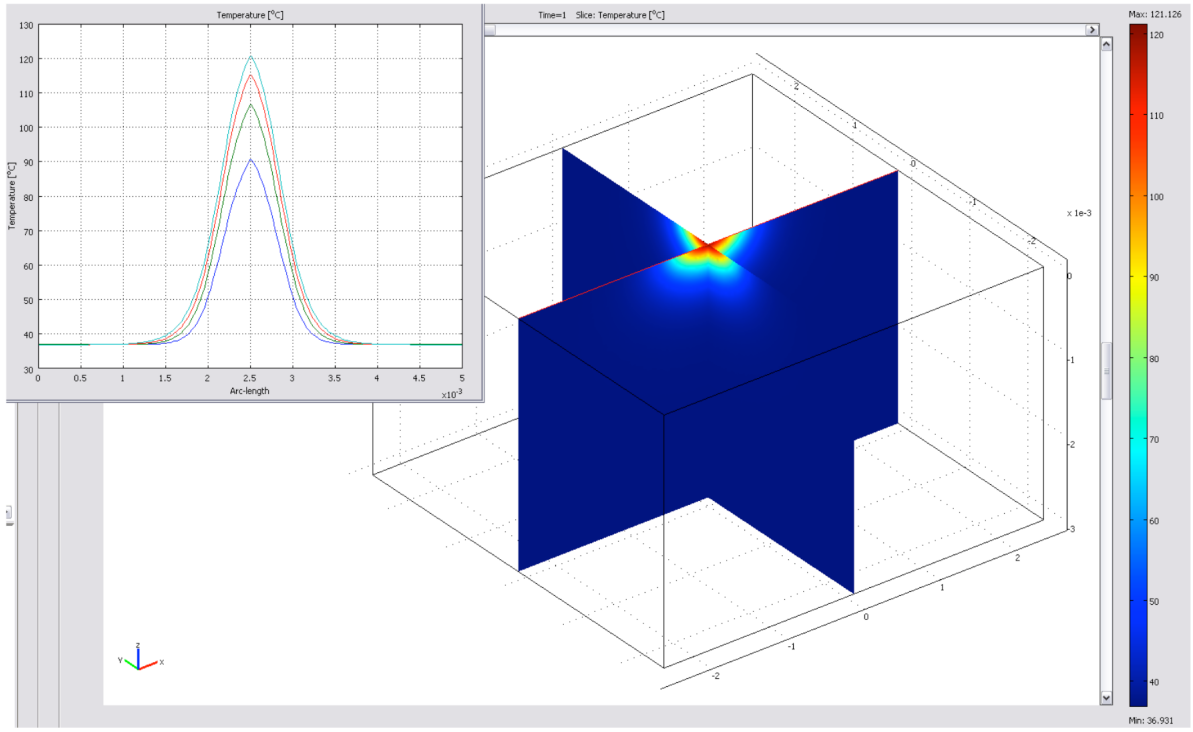

Figure 1. CO_2 laser source, temperature distribution, and penetration depth.

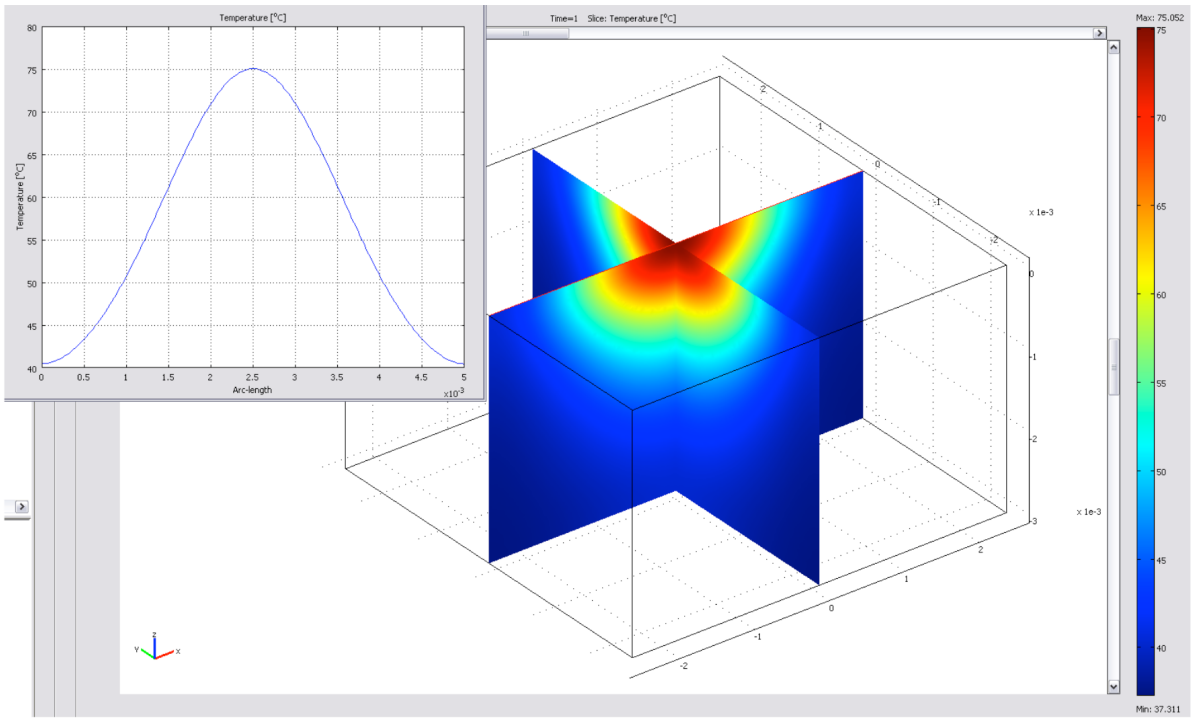

Figure 2. Nd:YAG laser source, temperature distribution, and penetration depth.

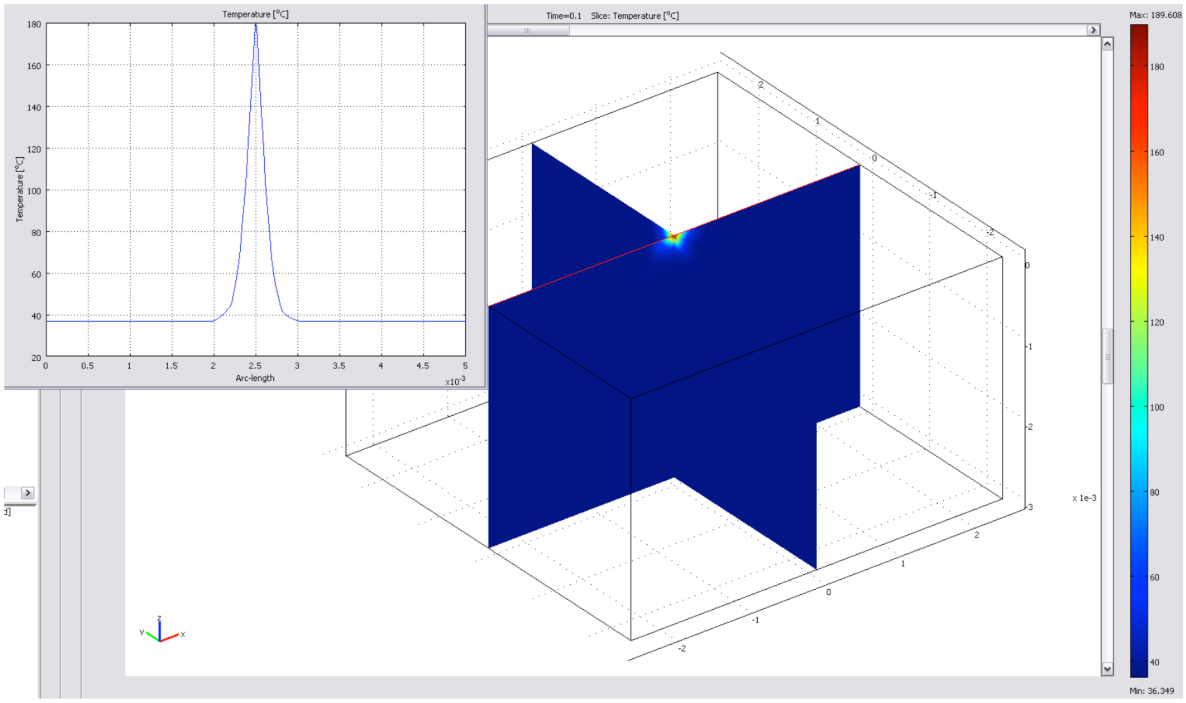

Figure 3. ArF excimer laser source, temperature distribution, and penetration depth.

the subject of tissue optical properties, published parameters display characteristic variations, especially in the infrared region. Optical properties of the skin layers show differences due to the random and inhomogeneous distribution of blood and pigments in skin. Considering these differences in optical and physical properties makes it possible to divide the skin tissue into sublayers. Absorption characteristics of hemoglobin, water, and lipids define the absorption of the whole skin tissue. Uniformly distributed spherical lipid droplets are defined to be the main scatterer elements in subcutaneous fat [13].

The light–tissue interaction is defined not only by the optical source parameters but also by the optical tissue parameters, such as the absorption, μ_a, coefficients that are dependent on wavelength. Towards the visible wavelengths, the absorption coefficient increases with a variation over several orders of magnitude such that it is in the range of 0.5–5 cm^{-1} at wavelengths of $\lambda < 625$ nm, whereas with $\lambda > 625$ nm in IR and near-IR regions, it varies between 0.01 and 0.5 cm^{-1} [14].

5. FEM simulations for multilayer tissue model

The mathematical modeling developed to obtain the solution for the bioheat equation has been implemented in order to analyze the dynamics of spatiotemporal temperature distribution $T(x, y, x, t)$ in biological tissues that are exposed to laser beam illumination. The skin is exposed to a continuous-wave HeNe laser with wavelength $\lambda = 633$ nm and Gaussian beam radius r_0. The skin tissue is modeled with various optical and thermal parameters obtained from different sources [1]. From the optical point of view, the selected model of the skin tissue is composed of six layers, whose ordering starts from the outside of the skin (the boundary between skin and air) and continues to the inside. Optothermal properties of the skin layers are summarized in Table 1 [1], and the geometry for laser–tissue interaction is shown in Figure 4 [1]. Figure 4 illustrates the multilayered structure of the skin model (not to scale for layer thicknesses), and it shows the interaction of laser and skin tissue in a 3-dimensional structure.

Table 1. Optical and thermal parameters for the multilayered model of skin.

Layer	d (mm)	α (cm^{-1})	ρ (g/cm^3)	c (J/g K)	$k\times 10^3$ (W/cm K)	$\omega_b \times 10^7$ (cm^3 blood/s g_{tissue})
Epidermis	0.06	25	1	3.77	2.09	0
Upper dermis	0.56	2.7	1	3.77	3.075	7.51
Blood plexus	0.1	25	1	3.77	3.075	7.51
Lower dermis	0.56	2.7	1	3.77	3.075	7.51
Subcutaneous fat	0.32	0.2	0.85	1.96	2.09	35.07
Muscle	> 2	11.2	1.05	3.94	6.42	4.509

In the simulation examples, initial temperature of the tissue was fixed at 36.6 °C and air temperature at 20 °C. The calculation volume of the skin tissue was chosen to be 5 mm × 5 mm × 4 mm, based on the estimation of the irradiated volume by the laser source (with radius $r_0 = 1$ mm). However, the multilayered structure of skin (as shown in Figure 4) presents certain limitations in the definition of the spatial step in the z direction, because of the minimal thickness of the layers, which is 0.06 mm and corresponds to the epidermis.

Simulation results on temperature distribution for the multilayer tissue model at different laser source powers and varying exposure durations are provided in Figures 5–11. In light of the results obtained from the simulations, it can be clearly stated that the proposed mathematical model enables the scientific prediction of thermal variations. Figures 5–11 show the temperature variation with tissue depth as a function of optical

power and also as a function of duration of exposure. When these simulation results are examined, the effects of skin layers with different optothermal properties, such as absorption coefficient, heat conductivity, and so on, can be observed. This is why the temperature distribution inside the tissue is usually a nonmonotonic function of depth.

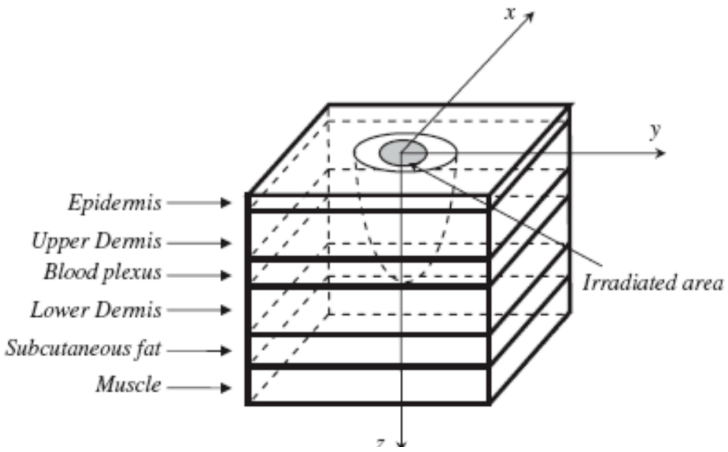

Figure 4. The multilayered model of skin and the geometry of laser–tissue interaction.

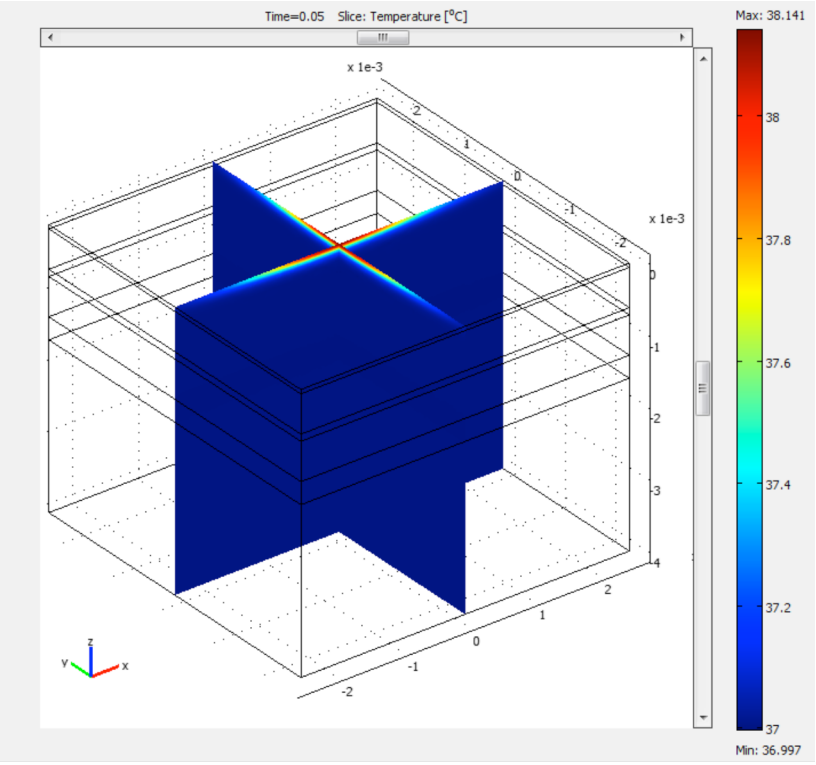

Figure 5. Temperature distribution and penetration depth for P = 50 mW, 50 ms duration.

All of the results from the above simulations of the multilayer model of human skin are summarized in Tables 2 and 3 and Figures 12 and 13 so as to clearly understand the variational effects of laser power and duration of

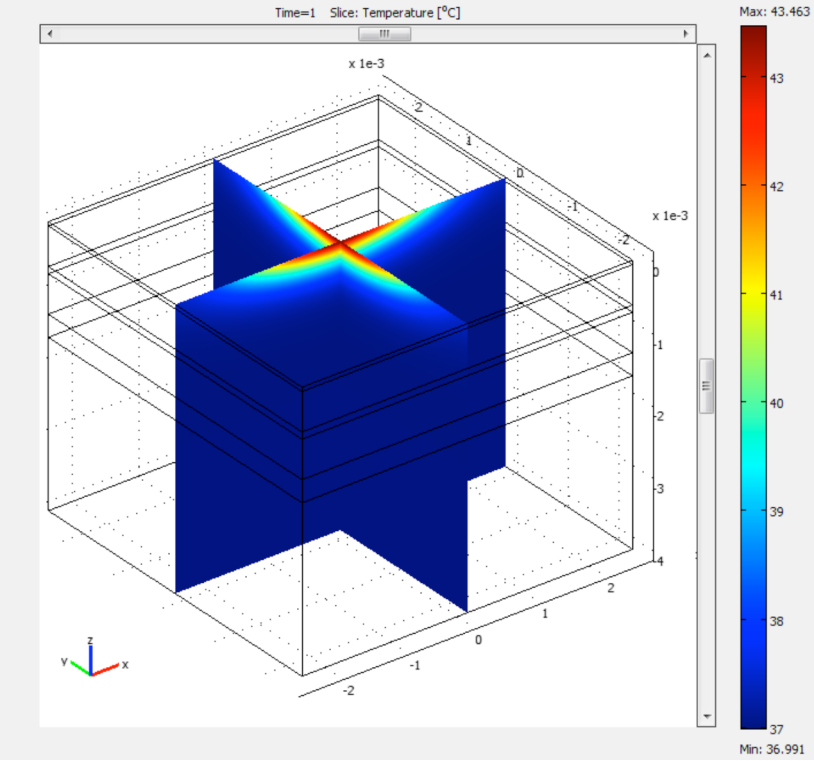

Figure 6. Temperature distribution and penetration depth for P = 50 mW, 1 s duration.

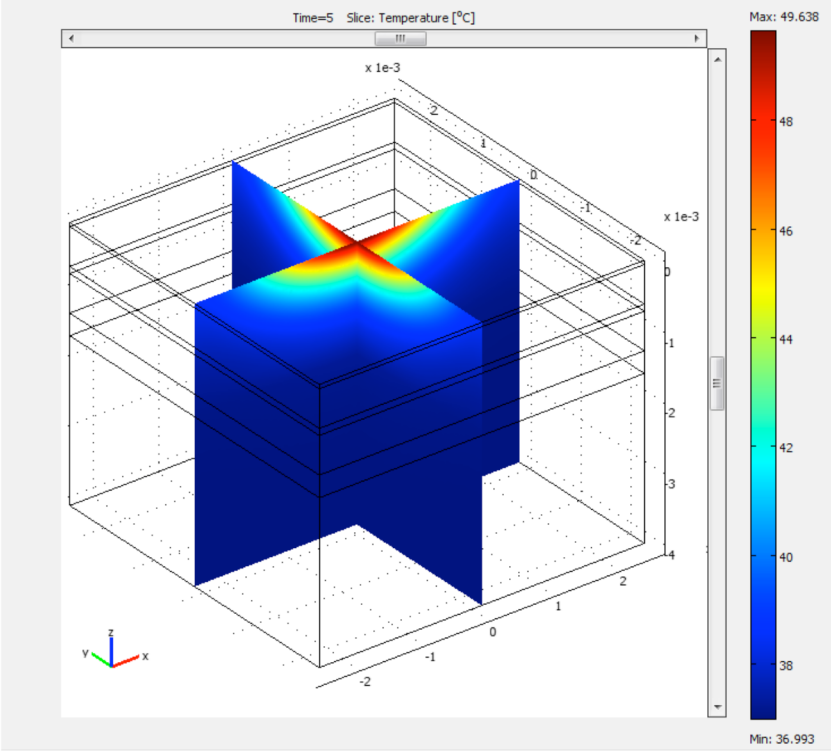

Figure 7. Temperature distribution and penetration depth for P = 50 mW, 5 s duration.

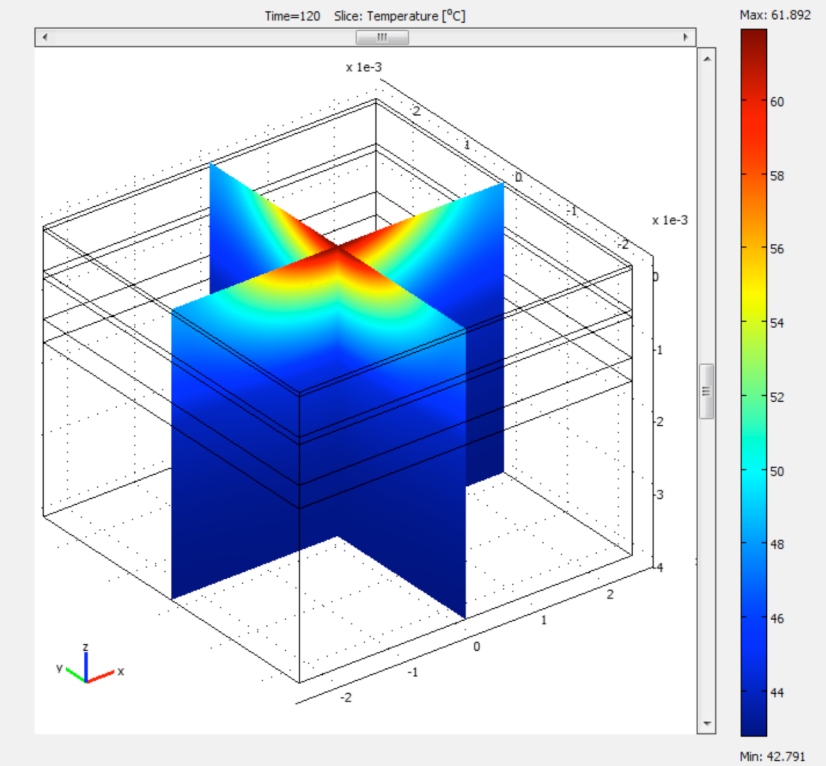

Figure 8. Temperature distribution and penetration depth for P = 50 mW, 120 s duration.

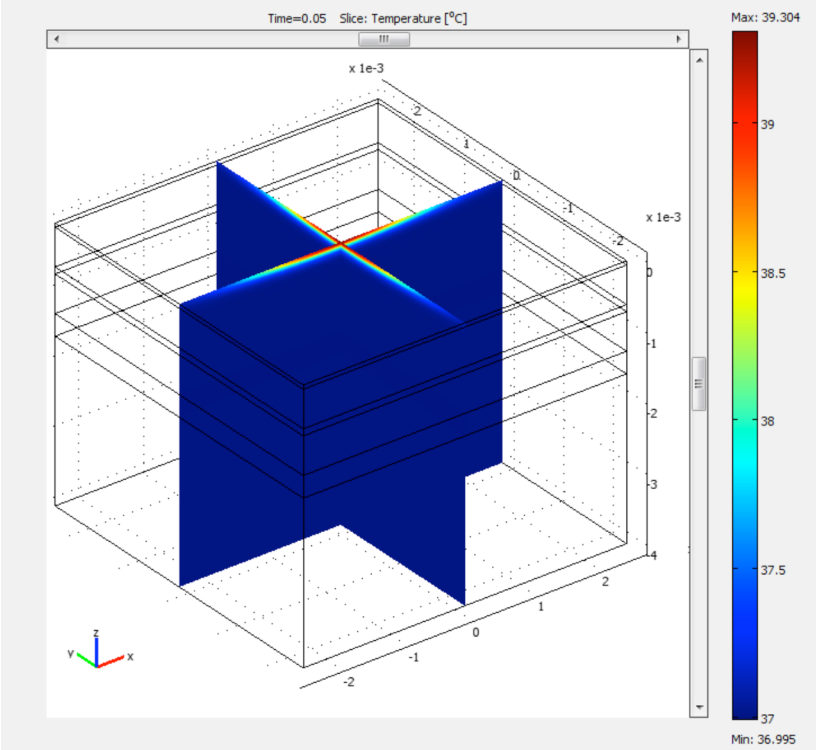

Figure 9. Temperature distribution and penetration depth for P = 100 mW, 50 ms duration.

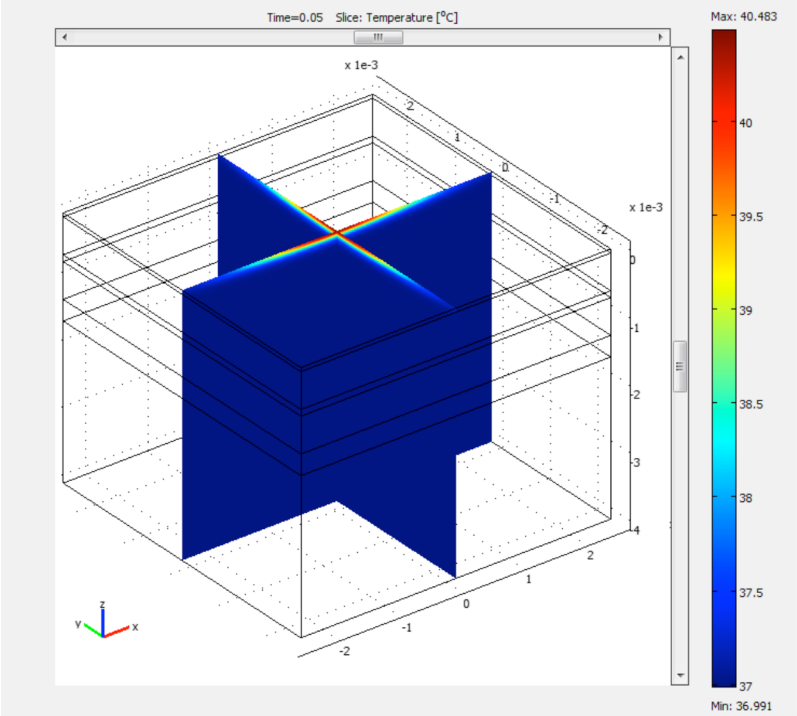

Figure 10. Temperature distribution and penetration depth for P = 150 mW, 50 ms duration.

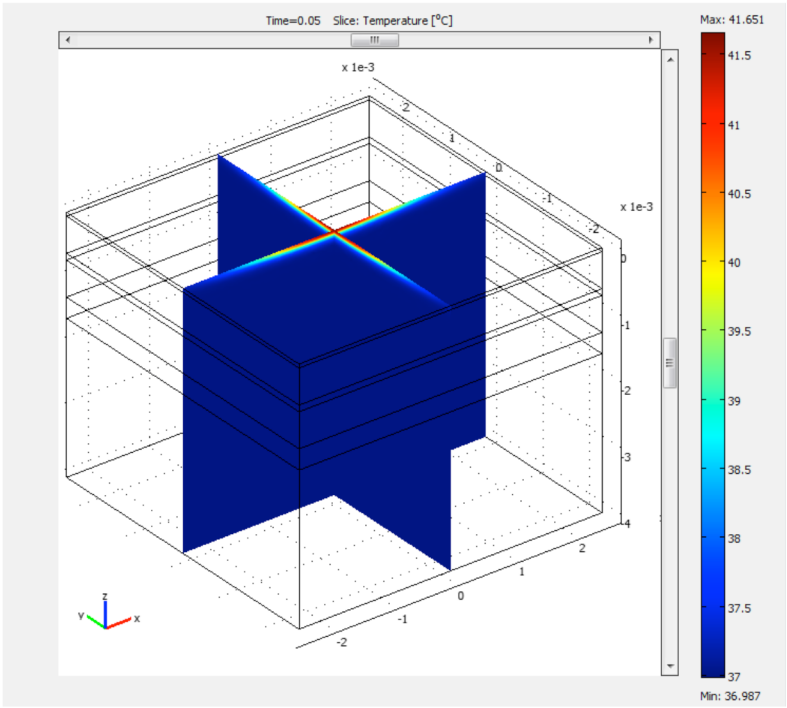

Figure 11. Temperature distribution and penetration depth for P = 200 mW, 50 ms duration.

exposure on the maximum temperature values. It can be clearly inferred that as duration of exposure to the laser is increased, maximum temperature induced on the skin increases. Additionally, as the laser power is increased by keeping the exposure duration constant, the maximum temperature due to exposure is increasing linearly.

Table 2. Maximum temperature values for varying laser power at exposure duration t = 0.05 s.

@ Duration of exposure t = 0.05 s				
Laser power (P) (mW)	50	100	150	200
Max. temperature (T) (°C)	38.141	39.304	40.483	41.651

Table 3. Maximum temperature values for varying exposure durations at laser power P = 50 mW.

@ Laser power P = 50 mW				
Duration of exposure (t) (s)	0.05	1	5	120
Max. temperature (T) (°C)	38.141	43.463	49.638	61.892

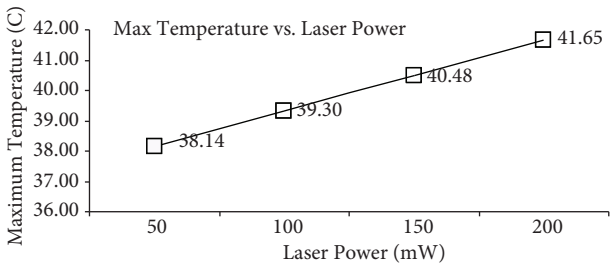

Figure 12. Effect of variation of laser power on maximum temperature @ exposure duration t = 0.05 s.

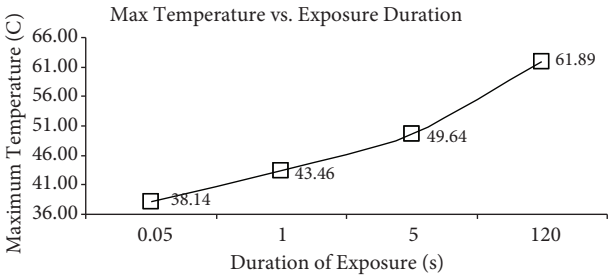

Figure 13. Effect of variation of exposure duration on maximum temperature @ laser power P = 50 mW.

6. Conclusions

In recent years, thermal analysis of biological tissues has attracted great attention owing to its significance in clinical and environmental sciences. In this respect, numerical modeling of interactions between light sources and living biological tissues was studied here. A new efficient numerical algorithm, the FEM, has been proposed to obtain the solution of the bioheat equation.

A single-layer human skin model was developed and simulated to predict the temperature distribution for laser light. In order to describe the heat flux flow, the Pennes bioheat equation was chosen, and the results were obtained numerically thanks to the FEM. These outcomes are in good agreement with the solutions from the literature that were obtained from established finite volume approaches.

The results obtained from this study reveal the important parameters whose effects are significant on the thermal response of skin, such as exposure duration and wavelength of the beam, as well as the area and type of tissue. For the single-layer model proposed, effects of different laser sources (CO_2, Nd:YAG, ArF excimer) with different absorption coefficient values at varying wavelengths were illustrated, and thermal distribution for the laser–tissue interaction was observed to be controlled by the wavelength of the laser beam and by the water absorption values of the skin.

To increase the reliability of the approach and for clearer representation of the physiological response of the skin tissue, further improvements to the analysis carried out in this study were also established with

multilayer modeling and simulation of the biological tissue to present a more realistic model. The numerical algorithm was investigated in detail, and the model was implemented to the irradiation of skin exposed to laser therapies.

In addition to the consistency of our proposed model, simulations also affirmed its satisfactory accuracy as a reference for clinical therapies. Laser light sources with different power outputs were also analyzed and the effect of varying exposure duration was investigated. These findings for the spatiotemporal temperature distribution, which is the result of optothermal treatment of tissue, will provide comparative advantages in selecting optimal laser treatment parameters in medical applications.

With the help of this study, some important contributions to the literature have been made on the topic of temperature predictions for biological bodies under exposure to light sources. As an initial step, usage of the FEM has extended the results of the FDTD approach, achieving better accuracy for complex geometries with less memory and disk space in shorter solution times. Moreover, simulation results for the single-layer model of human skin have provided a general and broad understanding of heat flux flow for different types of laser lights at various wavelengths. Not limited to this single-layer model, by proposing the multilayer human skin model, the reliability of the approach has been enhanced and it provided a better understanding for the response of skin to interaction with light sources. Effects of exposure duration, laser power, and the wavelength of the laser beam were investigated so that appropriate choice of parameters could be made for medical laser treatment.

Furthermore, the heating effect of several light sources on human skin tissue has not been reported yet, and the safety distances of lights have not yet been proposed, although there have been some efforts by the International Commission on Nonionizing Radiation Protection, the International Electrotechnical Commission, and the American National Standards Institute to develop regulations about light hazards concentrated not only on skin tissue but also on eye injury due to radiated energy. Consequently, the thermal analysis carried out in the present work can be used for this goal, as well.

Acknowledgment

This study was supported by the Boğaziçi University Research Fund with Project Code 5698.

References

[1] Fanjul-Velez F, Romanov OG, Arce-Diego JL. Efficient 3D numerical approach for temperature prediction in laser irradiated biological tissues. Comput Biol Med 2009; 39: 810-817.

[2] Pennes HH. Analysis of tissue and arterial blood temperatures in the resting human forearm. J Appl Physiol 1948; 1: 93-122.

[3] Yoon G, Welch AJ, Motamedi M, Gemert MV. Development and application of three-dimensional light distribution model for laser irradiated tissue. IEEE J Quantum Elect 1987; 23: 1721-1733.

[4] Vo-Dinh T. Biomedical Photonics Handbook. Boca Raton, FL, USA: CRC Press, 2003.

[5] Wang L, Jacques SL, Zheng L. MCML-Monte Carlo modeling of light transport in multi-layered tissues. Comput Meth Prog Bio 1995; 47: 131-146.

[6] Kolios MC, Worthington AE, Sherar MD, Hunt JW. Experimental evaluation of two simple thermal models using transient temperature analysis. Phys Med Biol 1998; 43: 3325-3340.

[7] Verhey JF, Mohammed Y, Ludwig A, Giese K. Implementation of a practical model for light and heat distribution using laser-induced thermotherapy near to a large vessel. Phys Med Biol 2003; 48: 3595-3610.

[8] Zhou J, Zhang Y, Chen JK. An axisymmetric dual-phase-lag bioheat model for laser heating of living tissues. Int J Therm Sci 2009; 48: 1477-1485.

[9] Ng EY, Tan HM, Ooi EH. Prediction and parametric analysis of thermal profiles within heated human skin using the boundary element method. Philos T Roy Soc A 2010; 368: 655-678.

[10] Citkaya AY, Seker SS. Modeling and simulation of temperature distribution in laser-tissue interaction. In: PIERS Progress In Electromagnetics Research Symposium; 20–23 March 2011; Marrakesh, Morocco. Cambridge, MA, USA: PIERS. pp. 844-847.

[11] Cheong WF, Prahl SA, Welch AJ. A review of the optical properties of biological tissues. IEEE J Quantum Elect 1990; 26: 2166-2185.

[12] Bashkatov AN, Genina EA, Kochubey VI, Tuchin VV. Optical properties of human skin, subcutaneous and mucous tissues in the wavelength range from 400 to 2000 nm. J Phys D Appl Phys 2005; 38: 2543-2555.

[13] Thompson MS. Photodynamic therapy utilizing interstitial light delivery combined with spectroscopic methods. PhD, Lund University, Sweden, 2004.

[14] Klose AD, Larsen EW. Light transport in biological tissue based on the simplified spherical harmonics equations. J Comput Phys 2006; 220: 441-470.