Effects of Memory Dependent Derivative of Bio-heat Model in Skin Tissue exposed to Laser Radiation

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Abstract

INTRODUCTION: Thermal processes are the essence of living organisms and are necessary for understanding life. The study of Transfer of heat in tissues is known as Bioheat transfer. Many techniques are developed for the thermal treatment of skin and other disease such as skin cancer, skin burns and injured skin tissue with laser. The tissue is inhomogeneous and at times anisotropic with complex thermal properties. Moreover, there may be skin tissue damage when irradiated with a laser beam.

OBJECTIVES: In this research a novel one-dimensional (1-D) bioheat model has been used with memory-dependent derivative (MDD) in Pennes’ bioheat transfer equation due to laser radiation and the thermal damage in tissue caused due to laser heating has been examined.

METHODS: Bioheat transfer model has been used with memory-dependent derivative (MDD) in Pennes’ bioheat transfer equation. The problem is solved using Laplace transform technique.

RESULTS: The temperature and thermal damage in the skin exposed to heating with laser radiation is calculated and obtained in physical form. The thermal reaction of skin tissues during laser radiation is studied under memory-dependent derivative (MDD) in Pennes’ bioheat transfer equation.

CONCLUSION: Analyzed a novel bioheat mathematical model on the basis of MDD involving time-delay parameter \( \chi \) for the Pennes’ bioheat transfer equation and applied to examine the thermal properties of the skin tissue for burns caused due to laser radiations. The thermal damages can be measured in a better way with the MDD model. The blood perfusion prevent the tissue damage by developing the cooling function. Effect of memory dependent derivatives and time delay parameter are represented graphically and analysed.

Keywords: Memory dependent derivative, Bioheat transfer, Time delay, Kernel function.

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Nomenclature

- \( T_b \): Blood temperature
- \( \omega_b \): Rate of blood perfusion
- \( t \): Time
- \( I_0 \): Intensity of the laser
- \( c \): Specific heat of the tissue
- \( c_b \): Specific heat of the blood
- \( \rho \): Tissue mass density
- \( k \): thermal conductivity of the tissue
- \( U(t) \): unit step function
- \( B \): factor of frequency
- \( \mu_a \): coefficient of absorption
- \( Q_{ext} \): heat generated per unit volume of tissues
- \( K(t-\xi) \): Kernel function
- \( C_1, C_2 \): Functions of diffuse reflectance \( R_d \)
- \( k_1, k_2 \): Functions of diffuse reflectance \( R_d \)
1. Introduction

Thermal processes are the essence of living organisms and are necessary for understanding life. Transfer of heat in tissues is complicated as it is affected by the flow of blood and asymmetrical structure of vascular tissues. The tissue is inhomogeneous and at times anisotropic with complex thermal properties. It also produces heat as part of the active metabolism. Maintenance of a fairly constant body temperature in a range of thermal environments implies that there is a continuous exchange of energy between deep, surface tissues and the environment.

Lipkin & Hardy [1] developed the method for measuring the temperature of human tissues. Giering[2] described the numerous thermal characteristics of biological tissues. Gardner et al. [3] proposed the 1-D light transfer case in tissue for fluence rate and escape function using Monte Carlo method. Cheng & Plewes [4] developed a technique to determine the thermal properties precise to the patient and its organ or tumor. Shrivastava and Vaughan [5] proposed a generic bioheat transfer model which include protection of energy, rate of transfer of heat from blood vessels to tissue and applicable to any type of tissue. Othman et al. [6] used the Penne’s bio-heat transfer equation for characterization of temperature variation in tissues.

Youn and Lee [7] evaluates the light dispersal and penetration depth in skin tissue using high-intensity light sources. Kengne et al. [8] predicts the distribution of temperature in a finite biological tissue with spatial heating and oscillatory surface. Deng & Liu [9] described the theoretical approaches to solve 3-D bioheat transfer problems in presence and absence of phase change. Kengne et al. [10] discussed the bio-heat transfer model for thermal traveling-wave dispersion in biological tissues. Kujawska [11] explored the ultrasonic method to determine the thermal conductivity of animal tissues using pulsed focused ultrasound. Kujawska et al. [11] developed a novel ultrasonic method to measure the thermal conductivity of certain animal tissues. Kumar et al. [12, 13] described the transfer of heat in a skin tissue of finite domain using metabolic heat. Sarkar [14] defined "the memory-dependent derivative in an integral form of a common derivative with a Kernel function on a slipping interval and insisted that this kind of definition is better than the fractional one for reflecting the memory effect (instantaneous change rate depends on the past state). Sarkar [14] pleaded the definition to be more intuitionistic for understanding the physical meaning and the corresponding memory dependent differential equation had more expressive force".

Youssef and Alghamdi [15] proposed a mathematical model one dimensional thermoelastic skin tissue of small thickness using dual-phase-lag heat law. Hobiny et al. [16] explored the bioheat model to thermal damage of living tissue caused by laser irradiation using the fractional order derivative. Ezzat et al. [17, 18] discussed the thermal responses of skin tissue using a fractional model of Bioheat equation with sinusoidal heat flux applied on the skin surface. Despite of this several researchers as Mahmoud et al. [19], Marin et al. [20, 21], Zhang and Fu [22], Abbas & Marin [23], Bhatti et al. [24], Marin [25, 26], Lata and Kaur [27-30], Riazet et al. [31] worked on different theory of thermoelasticity.

In this research a novel one-dimensional (1-D) bioheat model has been used with memory-dependent derivative (MDD) in Pennes’ bioheat transfer equation due to laser radiation and the thermal damage in tissue caused due to laser heating has been examined. Laplace transform technique is used to solve the given problem. The temperature and thermal damage in the skin exposed to heating with laser radiation is calculated and obtained in physical form. Effect of memory dependent derivatives and time delay parameter are represented graphically and analyzed.

2. Basic Equations

For the differentiable function f(t), Wang and Li [32] introduced the first-order MDD with respect to the time delay ξ > 0 for a fixed time t:

$$D_\chi f(t) = \frac{1}{\chi} \int_0^t K(t - \xi)f'(\xi)d\xi.$$  (1)

The Kr(t-ξ) and the time delay parameter ξ depends on the material properties. The kernel function K(t−ξ) is differentiable with respect to the variables t and ξ. Following Ezzat et al. [33-35] the kernel function K(t−ξ) is taken here in the form

$$K(t - \xi) = 1 - \frac{2b}{\chi} \frac{a^2}{\chi} (t - \xi)^2 =$$

$$\begin{cases}
1 & a = 0, b = 0 \\
1 + \left(\frac{\xi - t}{a}\right)^2 & a = 0, b = 1/2 \\
\frac{\xi - t + 1}{a} & a = 0, b = 1/\chi \\
\left(1 + \frac{\xi - t}{a}\right)^2 & a = 1, b = 1 \\
\end{cases}$$

where a and b are constants.

Following Sarkar [14] and Hobiny et al. [16] the memory dependent bio heat transfer equation with instant surface heating due to laser irradiation is given by

$$k\nabla^2 T = \left(1 + \chi D_\chi\right)\left(\rho C_T - Q - Q_{ext}\right).$$  (3)
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The temperature distribution in a semi-infinite biological tissue with instantaneous surface heating and with the laser thermal source of 1-D model of Memory-Dependent Derivative Pennes bioheat transfer equation (MDDPBE) in a finite medium is considered. The 1-D form of Eq. (3) by taking \( Q_m \) as constant, is written as:

\[
\frac{\partial^2 \theta}{\partial x^2} = (1 + \chi D) \left( \rho C \frac{\partial \theta}{\partial t} - \omega_b \rho_b c_b (T_b - \theta + T_0) - \frac{Q_m}{k} - \hat{I}_0 \hat{m} \left[ U(t) - U(t - \tau_p) \right] \right) - \frac{\rho \omega_b c_b k^2 e^{-\kappa_2 x}}{k}.
\]  

The initial conditions are:
\[
\theta(x,0) = \theta(x,0) = 0, 0 \leq x \leq L.
\]  

The skin tissue is exposed to instantaneous surface heating. We consider that heat flux \( \to 0 \) deep inside the tissue. Therefore, applicable boundary conditions are:
\[
\theta(0,t) = \theta_w \left. \frac{\partial \theta}{\partial x} \right|_{x=L} = 0, 0 \leq x \leq L, t > 0.
\]  

For simplify the solution, following non-dimensional quantities are given by
\[
T' = \frac{T - T_0}{T_b - T_0}, \quad T_b' = \frac{T_b - T_0}{T_b - T_0}, \quad t' = \frac{k}{\rho c \ell^2 t}, \quad \xi' = \frac{k}{\rho c \ell^2} x' = \frac{k}{\rho c \ell^2} \ell \xi, \quad t' = \frac{k}{\rho c \ell^2} t', \quad \xi' = \frac{k}{\rho c \ell^2} x', \quad t' = \frac{k}{\rho c \ell^2} t, \quad \xi' = \frac{k}{\rho c \ell^2} x.
\]  

Using these non-dimensional quantities defined in (15), the governing Equation (12) and initial and boundary conditions (13) and (14) can be written as (by ignoring dashes)
\[
\frac{\partial^2 \theta}{\partial x^2} = (1 + \chi D) \left( \rho C \frac{\partial \theta}{\partial t} - \frac{R_m}{k} - \frac{R_f}{U(t)} \right) - \frac{C_1 e^{-\kappa_1 x}}{k} - \frac{C_2 e^{-\kappa_2 x}}{k}.
\]  

Laplace transforms is given by
\[
\mathcal{L}[f(x,t)] = \int_0^\infty e^{-st} f(x,t) dt = \mathcal{F}(x,s),
\]  

with basic properties
\[
\mathcal{L} \left( \frac{\partial f}{\partial t} \right) = s \mathcal{F}(x,s) - f(x,0),
\]  

\[
\mathcal{L} \left( \frac{\partial^2 f}{\partial t^2} \right) = s^2 \mathcal{F}(x,s) - sf(x,0) - \frac{df}{dt} \bigg|_{t=0}.
\]  

\[
\mathcal{L} \left( \chi D \right) = G
\]  

\[
= \frac{\tau_0}{\chi} \left( 1 - e^{-s \tau_0} \right) \left( 1 - \frac{2b}{s \chi^2} + \frac{a^2}{s^2 \chi^2} \right) - \frac{b^2}{s^2 \chi^2} e^{-s \tau_0}.
\]  

\[
= \frac{\tau_0}{\chi} \left( 1 - e^{-s \tau_0} \right) \left( a = 0, b = 0 \right).
\]  

Thus we get
\[
D^2 \hat{\theta} = (1 + G) \left( \rho C \hat{\theta} - \frac{R_b T_b}{s} + R_b \hat{\theta} \right)
\]  

\[
= \frac{R_m}{s} - \frac{R_f}{s}, \quad \frac{1}{s} - e^{-s \tau_p} \left( C_1 e^{-\kappa_1 x} - C_2 e^{-\kappa_2 x} \right).
\]  

where
\[
D = \frac{d}{dx}.
\]  

which can be further simplified as
\[
(D^2 - \eta^2) \hat{\theta} = \chi_1 + \chi_2 e^{-\kappa_1 \xi} + \chi_3 e^{-\kappa_2 \xi}
\]  

Where
\[ \eta^2 = (1 + G)(\rho Cs + R_b) \]  
\[ \zeta_1 = -(1 + G)\left(\frac{R_b T_b}{s} + \frac{R_m}{s}\right) \]  
\[ \zeta_2 = -(1 + G)\left(\frac{R_b}{s}[1 - e^{-\gamma}]C_1\right) \]  
\[ \zeta_3 = -(1 + G)\left(\frac{R_b}{s}[1 - e^{-\gamma}]C_2\right) \]

And the boundary conditions after the application of Laplace transform (17) takes the form
\[ \bar{\theta}(0, s) = -\frac{\theta_0}{s}\int_0^d \frac{d\theta}{dx} \bigg|_{x=d} = 0.0 \leq x \leq d, Re(s) > 0. \]

By using the boundary conditions defined in Eq. (29), in Eq. (24), the exact solution is obtained as:
\[ \bar{\theta}(x, s) = \frac{\theta_0}{s}\cosh(\eta(x-d)) + \zeta^4 \frac{\cosh \eta x}{\cosh \eta d} \]
\[ \zeta_6 \frac{\sinh \eta x}{\eta} - \zeta_4 \frac{\eta^2 + \zeta_2 \eta + \zeta_2^2}{\eta^2} + \zeta_3 \frac{\eta^4 + \zeta_2 \eta^3 + \zeta_2^2 \eta^2}{\eta^4} \]

Where
\[ \zeta_6 = \frac{\zeta_4}{\eta^2} - \frac{k_2 \eta^2}{k_1^2 - \eta^2} - \frac{\zeta_3}{k_2^2 - \eta^2} \]
\[ \zeta_5 = \zeta_4 \frac{\eta^2 + \zeta_2 \eta + \zeta_2^2}{\eta^4} - \zeta_3 \frac{\eta^4 + \zeta_2 \eta^3 + \zeta_2^2 \eta^2}{\eta^4} \]

The thermal damage i.e. evaluation of burn caused by laser radiation, following Jasiński [37], Askarizadeh & Ahmadikia [38] is given by:
\[ \Omega = \int_0^\infty Re^{-\eta \tau} \theta d\tau. \]

4. Results and discussion

Following Askarizadeh & Ahmadikia [38] for numerical results, the specific values of different parameters are
\[ \rho_b = 1060 \text{kg/m}^3, \]
\[ c_b = 3860 \text{J/kg K}^{-1}, \]
\[ \omega_b = 1.87 \times 10^{-3} \text{s}^{-1}, \]
\[ T_b = 37^\circ \text{C}, \]
\[ Q_m = 1.19 \times 10^3 \text{W/m}^{-3}, \]
\[ g = 0.9. \]
\[ \mu_s = 12000 \text{m}^{-1}, \]
\[ R = 8.313 \text{J/mol K}, \]
\[ Ea = 6.28 \times 105 \text{J/mol}, \]
\[ c = 4187 \text{J/kg K}^{-1}, \]
\[ k = 1000 \text{kg m}^{-3}, \]
\[ \tau_p = 10s, \]
\[ L = 0.03 m, \]
\[ \mu_a = 40 \text{m}^{-1}, \]
\[ T_0 = 37^\circ \text{C}, \]
\[ B = 3.1 \times 1098 \text{s}^{-1}. \]

The results are simulated using MATLAB software and illustrated graphically. The impact of laser source on the skin surface was incorporated. The proposed mathematical models depends on the bio-heat transfer found and suitable boundary conditions. The conducting heat source, metabolic and perfusions are used in the formulations of the mathematical model. Numerical results are presented graphically in Figures 1–4 to study the influence of memory dependent derivative, kernel function K(t−ξ), the laser exposure time τp, the thermal relaxation time τ0 and the time delay parameter χ on the temperature and the thermal damages. The skin tissue is considered as .03 m thick and the reference temperature is taken equal to skin normal temperature, that is, T0 = Tb = 370C.

Figure 1 exhibits the deviation of temperature along the distance x by keeping the values of τ0 = 5s and τp = 10s with different values of memory dependent derivative, kernel function K(t−ξ). It is seen that the temperature start rising with the distance and decrease as the blood perfusions in skin increases. Figure 2 illustrates the thermal damage w.r.t. time t keeping the values of τ0 = 5s and τp = 10s with different values of memory dependent derivative, c The kernel function K(t−ξ) when a=0, b=0 has the highest effect on the thermal damages.
Figure 1. Temperature variations w.r.t. skin depth for different values of $K(t - \xi)$.

Figure 3 exhibits the variation of temperature along the distance $x$ by keeping the values of $\tau_0 = 5s$ and $\tau_p = 10s$ with different values time delay parameter $\chi$ for kernel function $K(t-\xi) = [1 + (\xi - t)/\chi]^2$ and $a=1$, $b=1$. It is seen that the temperature start from the utmost value and decrease rapidly.

Figure 2. The variation of thermal damage w.r.t. different values of $K(t - \xi)$. 
Figure 3. Temperature distributions w.r.t. skin depth for different values of time-delay $\chi$ at $K(t - \xi) = [1 + (\xi - t)/\chi]^{2}$.

Figure 4 demonstrates the thermal damage w.r.t. time $t$ keeping the values of $\tau_g = 5s$ and $\tau_p = 10s$ with different values time delay parameter $\chi$ for kernel function $K(t - \xi) = [1 + (\xi - t)/\chi]^{2}$ and $a=1$, $b=1$. It is seen that the thermal damage start from the utmost value and decrease rapidly. It is observed that $\chi = .009$ has the highest impact on the thermal damage.

Figure 4. The variation of thermal damage at skin surface with different values of time-delay $\chi$ and $K(t - \xi) = [1 + (\xi - t)/\chi]^{2}$. 
4. Results and discussion

The main objective of this research work is to analyze a novel bioheat mathematical model on the basis of MDD involving time-delay parameter $\chi$ for the Pennes’ bioheat transfer equation and applied to examine the thermal properties of the skin tissue for burns caused due to laser radiations. The thermal damages can be measured in a better way with the MDD model. The blood perfusion prevent the tissue damage by developing the cooling function. In this research, the memory-dependent derivative involving time-delay parameter $\chi$ becomes a new measure of efficiency for bioheat transfer in the skin tissues. These results may be beneficial in the study and further improvements in the applications of thermotherapy in skin tissues.

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