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Analysis of Thermal Damage in Biological Tissue during External Step Heating Using Non-Fourier Bio-Heat Transfer Model – A Finite Difference Approach

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Abstract

The present work focuses on the thermal damage in living tissue under an external step-heating exposure. A Non-Fourier type of bioheat transfer model, including the effect thermal relaxation time due to thermal inertia and microstructure of biological tissue, has been adopted to investigate the thermal damage. A trainsient blood perfusion rate has been taken at different locations of the body in this analysis. Considering all of the transient PDEs, the implicit Backward in Time and Central in Space (BTCS) framework has been used to create the necessary finite difference equations. Burn integral relation proposed by Henriques, has been undertaken to predict second-degree and third-degree burn time. Finally, a comparison is proposed for three different bioheat transfer models like Penne's, Thermal wave, and Dual-Phase Lag (DPL) models to illustrate the effect of different relaxation times on thermal damage.

Keywords: Thermal relaxation time; Dual-phase lag model; Henrique's burn model; Finite difference method.

1. Introduction

Heat transfer in the biological system is one of the burning topic from few decades among the researchers. It has vast application in the field of Biotechnology and Biothermomechanics. Various popular practical applications like the thermal treatment of cancer by electromagnetic hyperthermia, skin burn prediction, cryosurgery, ocular surgery, and many more. Analysis of heat transfer through the anisotropic system like biological tissue is challenging as various factors like blood perfusion, metabolic heat generation, heat absorption due to an external source, etc., are considered. In 1948, Harry H. Penne first proposed a biological heat transfer model to find the radial temperature distribution of the forearm. The main motive of his analysis is to evaluate heat transfer in the forearm in terms of local heat generation and blood flow rate inside the tissue [1]. Here blood flow rate (basically termed as Volumetric Perfusion Rate) is an essential factor for thermoregulation. Also, blood vessel size and location with respect to the skin have a firm standpoint in heat transfer in human skin [2, 3]. So, considering all these parameters, it is seen that Penne's bio-heat model is sufficient for demonstrating the heat transfer inside the skin as given as,

$$\rho c \frac{\partial T}{\partial t} = \nabla.(k\nabla T) + \omega_b \rho_b c_b (T_a - T) + Q_{met} + Q_{ext} \quad (1)$$

Where ρ, c, k signify density, specific heat, the thermal conductivity of the tissue respectively; ω_b is the blood perfusion rate; Q_{met} is heat generation due to metabolism, and Q_{ext} is heat generation due to external means of heat source. However, Penne's model is based on the classical Fourier's law of heat conduction. But there are very few situations where Penne's paradigm is not enough to evaluate temperature distribution. However, the assumption of heat propagation instantaneously throughout the body is reasonable for most real-life applications. But some exceptional practical applications where heat transfer behaves in a non-Fourier manner. In these individual cases, thermal disturbances propagate with a finite speed, according to Cattaneo and Vernott [4, 5]. They have independently modified Penne's equation and added τ_q to capture the effect of thermal wave behavior, which is underestimated in Fourier's theory. The following equation is the linear extension of Penne's equation, known as hyperbolic bioheat transfer equation, as it has two double derivatives (wave terms), which modifies the parabolic Fourier equation into the thermal wave equation,

$$T_q \rho c \frac{\partial^2 T}{\partial t^2} = \nabla.(k\nabla T) + \omega_b \rho_b c_b (T_a - T) + q_{met} + q_{ext} + T_{q_{met}} \frac{\partial q_{met}}{\partial t} + T_{q_{ext}} \frac{\partial q_{ext}}{\partial t} - (\tau_q \omega_b \rho_b c_b + \rho c) \frac{\partial T}{\partial t} \quad (2)$$

Here T_q is known as thermal relaxation time, which depends on thermal diffusivity (α) and the speed of the thermal wave in the particular medium (C_t) [6]. This is very similar to the sound wave; that's why the propagation of thermal disturbances can be termed as "Second Sound Wave" [7]. Most research in bioheat transfer has been done in a numerical or analytical approach, as in vivo experimental analysis is very expensive, risky, and challenging. Some extensive work has been done to find the value of phase lag due to thermal inertia for biological tissues in various situations. It has been observed that the thermal relaxation time of biological materials is much larger

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than any other engineering materials due to their heavy anisotropic nature [6,8,9]. However, many works are searching for the values of thermal relaxation time (T_q) due to the lack of data in bioheat transfer field. It is further modified by considering the microstructural interaction for a fast time-dependent heat transfer process. This concept is absent in the thermal wave model bioheat transfer. Considering the phase lag due to temperature gradient (T_t), the following equation has been developed and also known as the Dual-Phase-Lag model of bioheat transfer,

$$q(x, t + T_q) = -k\nabla T(x, t + T_t) \qquad (3)$$

T_q and T_t are the relaxation time due to "thermal inertia" and "microstructural interaction," respectively. T_q specifies the phase lag due to itertial effect of heat flux associated with conductive heat transfer through the medium. In contrast, T_t is the phase lag due to temperature gradient across the medium caused by microstructural diversity. A massive amount of work is going on to find out the values of both phase lags in the analytical or numerical approach (few experimental). However, there is still a lack of data for thermal phase lags; extensive analysis is being required to evaluate the phase lag times in different situations.

Now evaluation of temperature distribution alone cannot satisfy the need for analysis until the amount of thermal damage of the biological tissue has been calculated. When the temperature of the biological body rises above 44^0C, the initiation of thermal damage occurs. This approach has been known as the "Critical Thermal Load" approach, that formulated such that the total burn is the function of simultaneously applied thermal loads. However, very few researchers have questioned it. Later Henriques and Moritz have proposed a model that represents skin burn as a chemical reaction, and based on the first-order Arrhenius equation, they proposed Arrhenius burn integration [10, 11]. This model calculates the thermal damage based on protein degradation and exposure time (t) at a particular absolute temperature [11]. Henrique's method has been widely used in numerical calculation as it is easy to implement in computer code. Though some other models are also available to predict the skin burn time, the most preferable is Henriques Burn Integral in this analysis as the advantages are observed in various works.

The present work focuses on the thermal damage in living tissue under an external step-heating exposure. Various effects of phase lag due to thermal inertia and biological tissue microstructure have been adopted to investigate the thermal damage. Also, temperature-dependent blood perfusion rate has been taken at different locations of the body in this analysis. Henrique's burn integral relation has been applied to calculate burn time for second-degree and third-degree damage. This study's main motive is to compare three bioheat transfer models like Penne's, Thermal wave, and Dual-Phase Lag (DPL) bioheat transfer models to illustrate the effect of varying relaxation times on thermal damage to make the analysis more realistic.

2. Mathematical formulation and numerical modeling

A 1D DPL bioheat model (equation 1) is used to evaluate the temporal variation of temperature in the living tissues. A temporal blood perfusion rate is taken in this analysis for two different body locations (arms and legs). An age-dependent metabolism rate (Q_m) has also been included in this numerical study. As shown in Fig. 1 the schematic diagram of the undertaken numerical model. Besides, the subsequent assumptions are taken in this analysis:

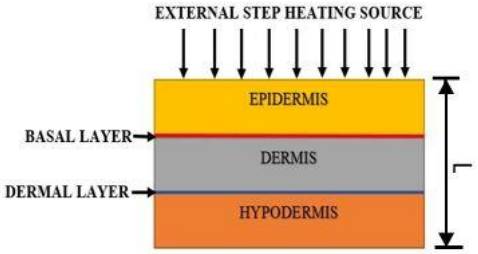

Fig. 1. Schematic diagram of the present model

- Heat absorbed due to protein denaturation of the cell is neglected.
- All the transport and thermodynamic properties are taken as constant.
- Sweating effect is neglected.

2.1. Initial and boundary conditions

As initial temperature distribution through the layers of the skin (epidermis, dermis, and hypodermis) does not affect the result [12], a constant initial temperature is assumed for the skin layers.

2.1.1. Initial Conditions

An external step heating flux of 45 kW/m^2 is applied to the skin for 20 s, followed by 30s cooling. The core body temperature is assumed as 310.15 K.

i. $T(x, 0) = 307.15\ K$; $(0 \le x < L)$

ii. $\frac{\partial T(x,0)}{\partial X} = 0$; $(0 \le x < L)$

2.1.2. Boundary Conditions

i. $Q(0,t) = -k\frac{\partial T}{\partial x}$; (0≤t≤a) (5a)

Where, $Q(0,t) = q_0[u(t) - u(t-a)]$

u(t) is a step function, and "a" denotes the duration of heating.

ii. $T(\mathrm{L},t) = 310.15\ K$; (5b)

2.2. Calculation of thermal damage using skin burn model

Henrique's burn integral equation [10] is used to calculate the extent of thermal impairment (1st, 2nd, and 3rd degree) in living tissues.It can be expressed like,

$$\Omega = \int_0^t P \exp(-E_a/RT)\ dt \qquad (6)$$

Table 1. The Severity of Thermal Damage [13]

Degree of burn	Value of integral	Skin layers	Severeness
1st degree	Ω = 0.53	Basal layer	Less severe
2nd degree	Ω = 1	Basal layer	More severe
3rd degree		Dermal layer	Most severe

Where P is a material parameter that corresponds to a frequency factor, R is the universal gas constant (8.314 J/mol K), E_a is the activation energy, and T signifies the temperature data required to calculate the thermal damage. Whereas the constants P and E_a have been calculated experimentally. The degree of burn has been calculated based on the following criteria [13]. This model is used to predict thermal damage of the biological tissues, which starts above 44^0C temperature. The level of the severity of burn can be predicted in terms of 1st, 2nd, and 3rd degree, which is explained in Table 1, and the values of P and $\frac{\Delta E}{R}$ are given in Table 2[14].

Table 2. The values of different parameters used in the skin burn model [14]

Temperature Range	Pre-exponential factor, P (1/s)	The ratio of activation energy to the gas constant, $\frac{\Delta E}{R}$ (K)
$44^0C \leq T < 50^0C$	2.185×10^{124}	93534.9
$T \geq 50^0C$	1.823×10^{51}	39109.8

2.3. Solution methodology

Here, the DPL bioheat transfer equation is discretized with a BTCS scheme and finite-difference method. The Thomas Algorithm is used to solve all of the linear algebraic equations (TDMA). After a successful grid independence test, a computer programme has been created in the MATLAB 2016 IDE with uniform step sizes for space and time of 0.00001 m and 0.01 s, respectively. According to Table 3, all of the thermophysical characteristics of skin and blood [15] are included. Additionally, figures for the arms and legs' basal perfusion rates [16] are used from Table number 4.

Table 3. Thermo-physical properties of skin and blood [15]

Layers	Thermal conductivity k (W/m K)	Density ρ (kg/m^3)	Specific heat C (J/kg K)	Thickness (m)
Epidermis	0.24	1200	3590	8×10^{-5}
Dermis	0.45	1200	3300	0.002
Hypodermis	0.19	1000	2500	0.01
Blood	--	1060	3770	--

Table 4. Values of basal blood perfusion rate [16]

Body position	Basal blood perfusion rate (m^3/s m^3)
Arms	0.00096154
Legs	0.001686

2.4. Numerical model validation

By taking into account all the transport and thermodynamic properties of skin and blood as well as the geometry of the solution domain as per the study of Udayraj et al. [15], there is an excellent agreement with an error of less than 1% between the result obtained from developed numerical model and the result obtained by Udayraj et al. [15] as shown in Fig. 3.

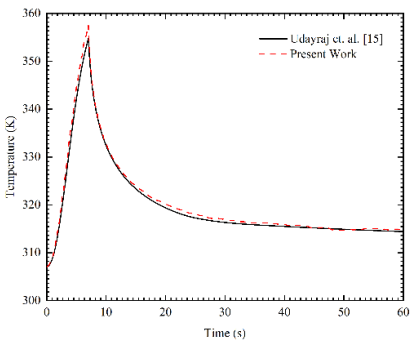

Fig. 2. Temporal temperature profile comparison with the findings of the literature [15]

3. Results and discussion

3.1 Diversity among the various established Bioheat transfer models on thermal damage

As a thermal safety personal experiences low, medium, and high-intensity exposures, it is essential to determine the most accurate time for thermal damage. In our current numerical analysis, three different bioheat transfer models, i.e., Penne's bioheat model (PBHT), Cattaneo's bioheat transfer model with finite velocity of thermal wave (TWMBHT), Dual-phase lag bioheat transfer model (DPLBHT), as shown in Fig.3, are compared based on a different degree of burning/thermal damage. It is observed from the analysis that the PBHT model records the least thermal damage time followed by DPL and TWMBHT.PBHT shows the least time as the heat propagation speed is infinite, which suddenly raises the basal and dermal cell temperature and drops it during cooling. TWMBHT records the highest burning time due to its wavy nature of heat propagation, where thermal damage recorded by the DPL model is less than TWMBHT because the diffusive nature of the heat transfer is dominated over the wave nature. Third-degree burning for the leg could not be reached for the stipulated heating duration. No significant difference in thermal damage between arms/hand and leg could be observed except the third-degree burning time for leg is recorded as 28.34s which is 2.64s more than arms.

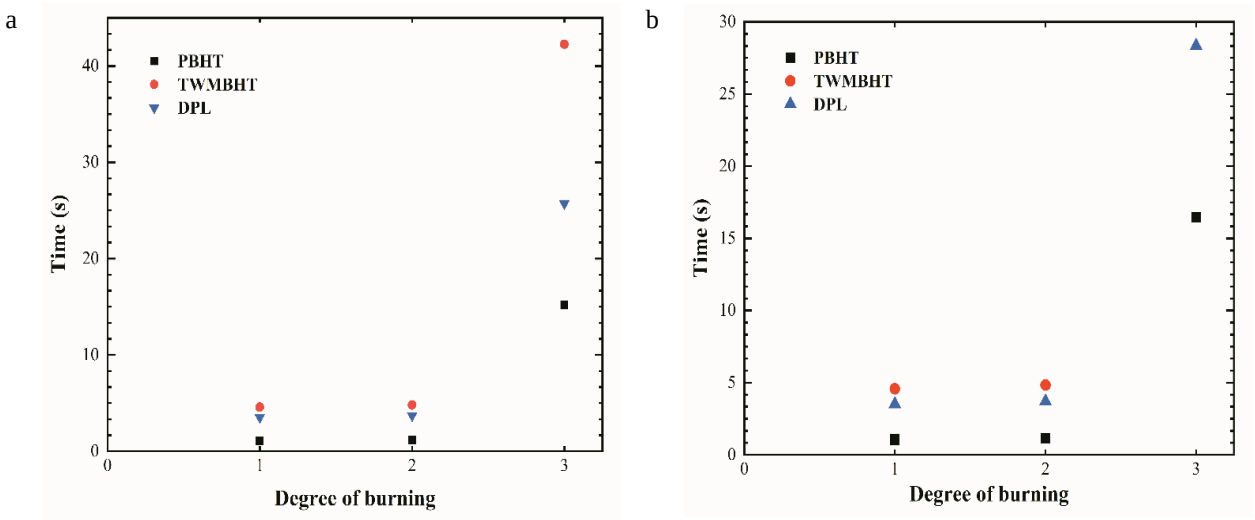

Fig. 3. Assessment of various bioheat transfer models for different degrees of burning times at two different body locations, i.e. (a) arms (b) legs

3.2 Effect of phase lag parameters on temperature distribution in tissues

Thermal relaxation time due to inertial effect of heat flux and temperature gradient mostly depend upon tissues, blood vessels, and different body locations. Fig. 4 and 5 show the temporal variation of temperature profile at the basal and dermal layers for two different body locations like arms and legs. Fourier's bioheat model (Penne's model) shows a abrupt rise and drop in temperature during the heating and cooling phase.DPL model having $T_t=T_q=0.1$ exhibits nearly the same temperature rise as PBHT model at both the locations (basal and dermal layers). The slope of the temporal rise in temperature during heating increase when the T_t value increases for the condition of $T_t>T_q$ due to the dominated diffusive nature of the heat propagation. The least temperature rise is recorded for the wavy nature dominated ($T_q>T_t$) heat propagation. Temperature shows a constant trend during the heating duration at the dermal layer for $T_q>T_t$, as depicted in Fig. 4(b) and Fig. 5(b). Still, the temperature curve increases gradually after the heating duration due to the slow penetration of the heatwave. All the temperature curves show a decreasing trend during the cooling phase, and finally, flatten attaining a steady state. It can also be noticed that the rate of temperature drop after switching off the heat source is less in the case of more T_t, as the heat will penetrate more depth before converging to the steady-state due to the dominated diffusive nature. The temperature rises recorded at basal and dermal layers is more in arms than legs. Maximum temperature rise at dermal layer is nearly identical for $T_q>T_t$ for both the location, but Penne's model and the model having dominated diffusive nature (T_q= 0 and T_t=10) show a higher temperature in arms than legs as presented in Fig. 4(b) and Fig. 5(b). It could also be observed that the maximum difference between the PBHT model and DPL model having more microstructural influence (T_q= 0 and T_t=10) is less at the dermal layer in legs than arms in terms of temperature data, but the result is reversed at the basal layer.

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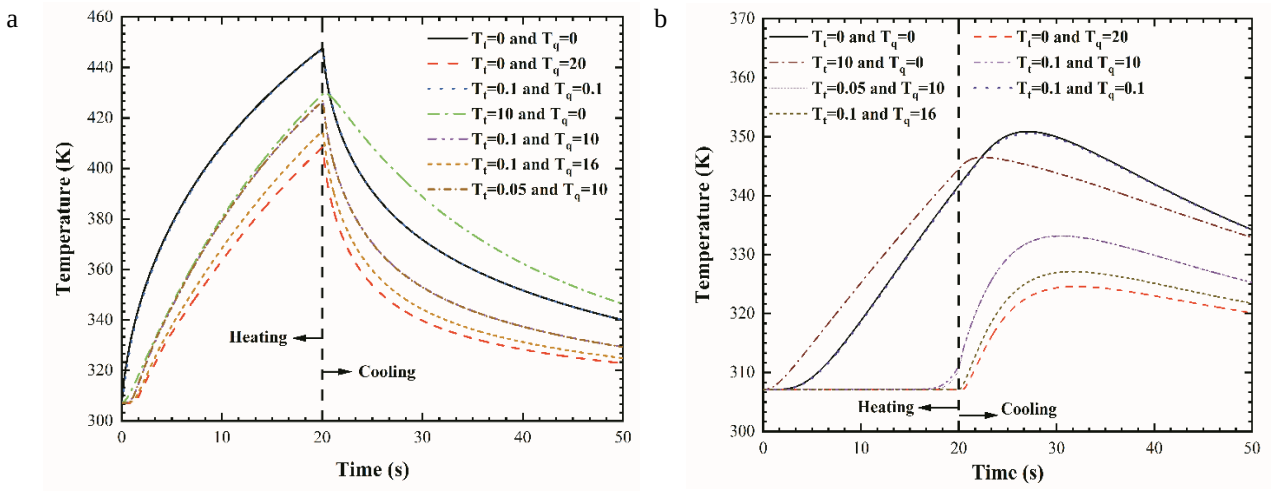

Fig. 4. Temporal temperature distribution at (a) basal layer and (b) dermal layer in arms for different phase lag times.

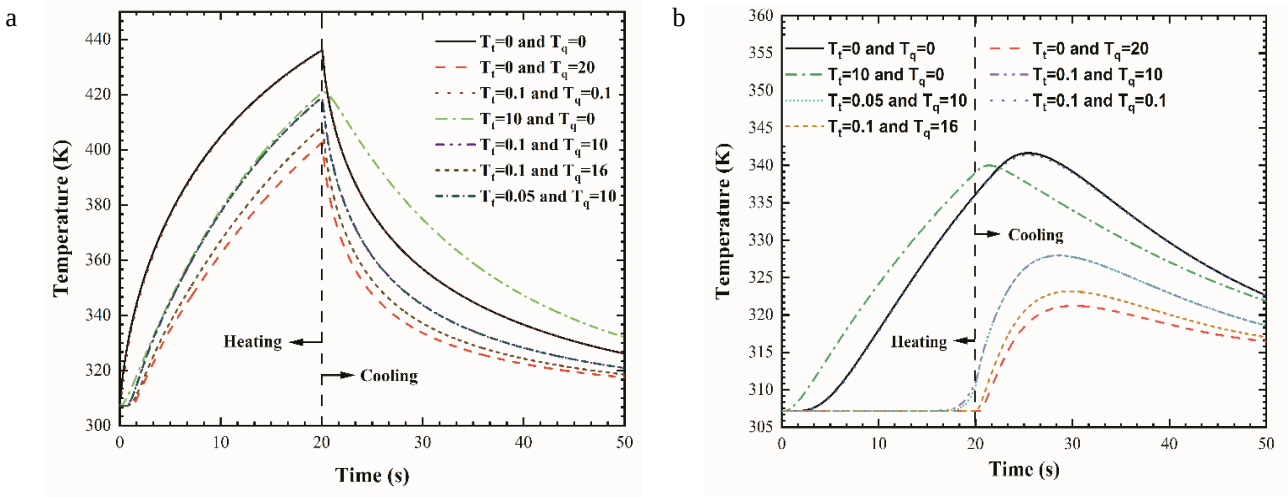

Fig. 5. Temporal temperature distribution at (a) basal layer and (b) dermal layer in the leg for different phase lag times.

3.3 Effect of relaxation times on thermal damage in tissues

As temperature distribution throughout the skin layer depends mainly on the values of relaxation times, it is essential to determine the 1st, 2nd, and 3rd-degree thermal damage at different layers of skin based on their severity. Thermal damage depends exponentially on the local tissue temperature. It also maintains a nonlinear relationship with different relaxation times.1st and 2nd-degree thermal damage could be evaluated at the basal layer when the skin burn factor becomes 0.53 and 1, respectively. It can be interpreted clearly from Fig. 6 and 7 that burning time decreases when T_t dominates. For a specific value of T_q, burning time decreases when the T_t value increases, as heat propagates faster due to its diffusive nature and rapidly increases tissue temperature. It could also be noticed that thermal inertia-dominated heat propagation takes more time to penetrate through the living tissue due to its wave nature; as a result, heat will reach a particular location of the skin after a long time due to its slow penetration nature. So thermal damage time increases if T_q increases for a particular value of T_t. There is no significant difference between time differences for thermal damage at basal layer (1st and 2nd degree) in legs and arms, but at the dermal layer, arms tissue is affected earlier than legs, as depicted in Fig.7. It could be said clearly from Fig.7 that 3rd-degree burning time is delayed due to the increment of T_q values; as a result of thermal damage curve sometimes looks like "S" shape (T_q=20, Tt=0 in arms as shown in Fig. 5a) and sometimes 3rd-degree burn could not be reached for the specific exposure time (T_q =20, Tt=0 and T_q =16, T_t=0.1 in legs as shown in Fig. 7(b)). PBHT model takes the least time taken for thermal damage at basal layer, but at the dermal layer, diffusive dominated single-phase lag (SPL) (T_t=10 and T_q=0) model damages tissue earlier than PBHT model due to its more penetrating nature.

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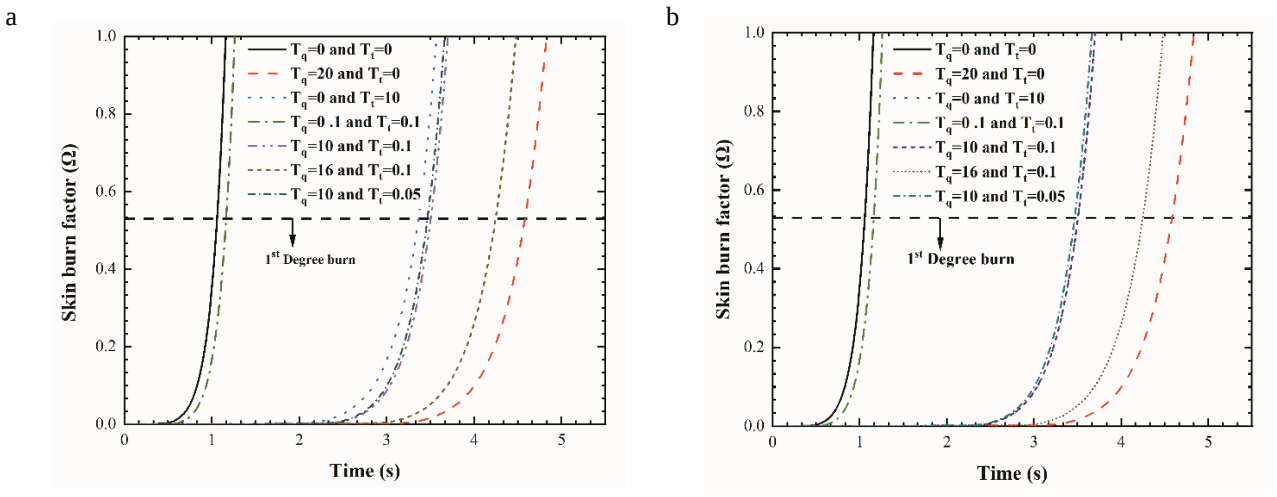

Fig. 6. Variation of skin burn factor with time at basal layer (1st and 2nd-degree thermal damage) for (a) arms and (b) legs

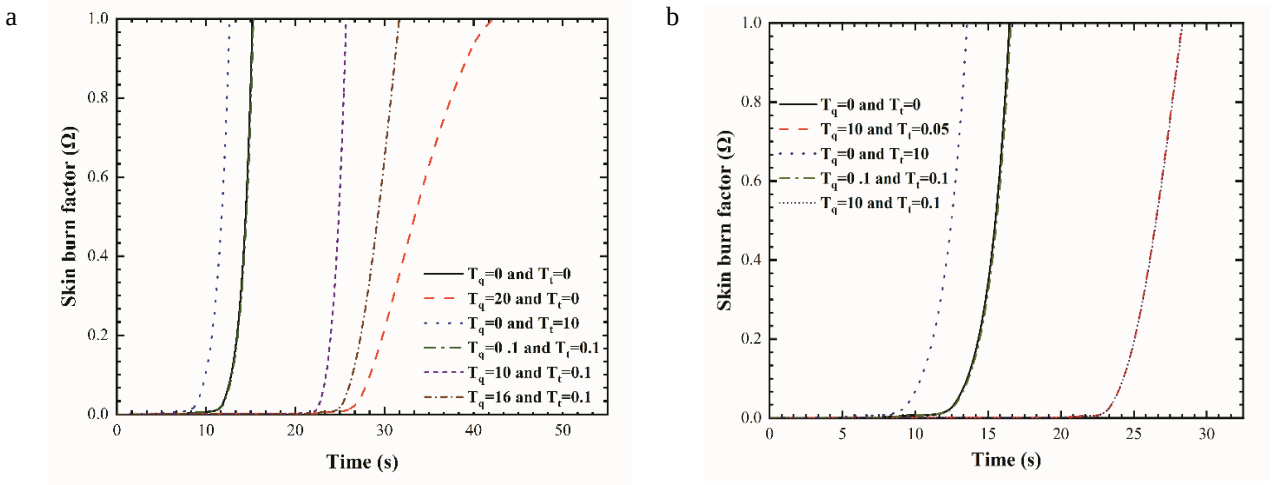

Fig. 7. Variation of skin burn factor with time at dermal layer (3rd-degree thermal damage) for (a) arms and (b) legs

4. Conclusion

Numerical analysis is conducted to observe the effect of different bioheat models (PBHT, TWMBHT, and DPL) on thermal damage at basal and dermal layers for two different parts of the human body, i.e., arms and legs when the skin is exposed by an external step heating flux having 45 kW/m^2 for 20s heating followed by 30s cooling. Thermal damage time at different layers is found to be more in the case of TWMBHT, followed by DPL and PBHT models. Temperature rise is found to be more in tissues when diffusive nature dominates the wavy nature of heat propagation. Different degrees of thermal damage is found at different layers. It could be observed that tissues are damaged earlier when relaxation time for microstructural interaction increases, but the wavy nature of the heat propagation delays the thermal damage time. At basal layers, thermal damage times for arms and legs are nearly the same, but tissues in arms are affected earlier than legs at the dermal layer.

References

(1) Pennes, H. H. (1948). Analysis of tissue and arterial blood temperatures in the resting human forearm. *Journal of applied physiology*, *1*(2), 93-122. https://doi.org/10.1152/jappl.1948.1.2.93

(2) Chato, J. C. (1980). Heat transfer to blood vessels. *Journal of biomechanical engineering*, *102*(2), 110-118. https://doi.org/10.1115/1.3138205

(3) Weinbaum, S. J. L. M., Jiji, L. M., & Lemons, D. E. (1984). Theory and experiment for the effect of vascular microstructure on surface tissue heat transfer—Part I: Anatomical foundation and model conceptualization. https://doi.org/10.1115/1.3138501

(4) Cattaneo, C. (1958). A form of heat-conduction equations which eliminates the paradox of instantaneous propagation. *Comptes Rendus*, *247*, 431.

(5) Vernotte, P. (1958). Les paradoxes de la theorie continue de l'equation de la chaleur. *Compt. Rendu*, *246*, 3154-3155.

(6) Mitra, K., Kumar, S., Vedevarz, A., & Moallemi, M. K. (1995). Experimental evidence of hyperbolic heat conduction in processed meat. https://doi.org/10.1115/1.2822615

(7) Chester, M. (1963). Second sound in solids. *Physical Review*, *131*(5), 2013. https://doi.org/10.1103/PhysRev.131.2013

(8) Vedavarz, A., Kumar, S., & Moallemi, M. K. (1994). Significance of non-Fourier heat waves in conduction. https://doi.org/10.1115/1.2910859

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*E-mail address:*ajnana1993@gmail.com

(9) Kaminski, W.: Hyperbolic heat conduction equation for materials with a nonhomogeneous inner structure. J. Heat Transf. Trans. ASME 112(3), 555–560 (1990). https://doi.org/10.1115/1.2910422

(10) Diller, K. R., & Hayes, L. J. (1991). Analysis of tissue injury by burning: comparison of in situ and skin flap models. *International journal of heat and mass transfer*, *34*(6), 1393-1406. https://doi.org/10.1016/0017-9310(91)90283-K

(11) Xu, F., Lu, T. J., & Seffen, K. A. (2008). Biothermomechanical behavior of skin tissue. *Acta Mechanica Sinica*, *24*(1), 1-23. https://doi.org/10.1007/s10409-007-0128-8

(12) S.C. Jiang, N. Ma, H.J. Li, X.X. Zhang, Effects of thermal properties and geometrical dimensions on skin burn injuries, Burns 28 (2002) 713–717. https://doi.org/10.1016/S0305-4179(02)00104-3

(13) Talukdar, P., Das, A., & Alagirusamy, R. (2016). Heat and mass transfer through thermal protective clothing–A review. *International Journal of Thermal Sciences*, *106*, 32-56. https://doi.org/10.1016/j.ijthermalsci.2016.03.006

(14) John A. Weaver, Alice M. Stoll, Mathematical model of skin exposed to thermal radiation, Aerospace Medical Research Department, Naval Air Systems Command, NADC-MR-6708, 1967. https://doi.org/10.21236/ad0659973

(15) Talukdar, P., Alagirusamy, R., & Das, A. (2014). Heat transfer analysis and second degree burn prediction in human skin exposed to flame and radiant heat using dual phase lag phenomenon. International Journal of Heat and Mass Transfer, 78, 1068-1079. https://doi.org/10.1016/j.ijheatmasstransfer.2014.07.073

(16) Shitzer, A., Arens, E., & Zhang, H. (2016). Compilation of basal metabolic and blood perfusion rates in various multi-compartment, whole-body thermoregulation models. International journal of biometeorology, 60(7), 1051-1064. https://doi.org/10.1007/s00484-015-1096-5