$$\tau_q \frac{\partial^2 q}{\partial t^2} + \frac{\partial q}{\partial t} = \alpha_s \frac{\partial^2 q}{\partial x^2} + \alpha_s \tau_T \frac{\partial^3 q}{\partial t \partial x^2} - \alpha_s \frac{\partial Q_L}{\partial x} + \frac{G\alpha_s}{(1-\varepsilon)} \frac{\partial T}{\partial x} + \frac{G\alpha_s \tau_T}{(1-\varepsilon)} \frac{\partial^2 T}{\partial t \partial x}$$

For highly absorbed tissues, the laser heating is approximated as the boundary condition of the second kind. The laser volumetric heat source or laser irradiance, Q_L, is zero and the boundary conditions are given by [10] the following.

$$q = \phi_{in}(1 - R_d) \qquad \text{for } x = 0 \text{ when } 0 < t < \tau_L \tag{9}$$

$$q = 0 \qquad \text{for } x = L \text{ when } 0 < t < \tau_L \tag{10}$$

where τ_L is the laser exposure time, φ_{in} is the incident laser irradiance, and R_d is the diffuse reflectance of light at the irradiated surface.

For strongly scattering tissues, laser heating is considered as a body heat source ($Q_L \neq 0$) but the irradiated surface is thermally insulated. The boundary conditions in this case can be represented as follows.

$$q = 0 \qquad \text{for } x = 0 \text{ when } 0 < t < \tau_L \tag{11}$$

$$q = 0 \qquad \text{for } x = L \text{ when } 0 < t < \tau_L \tag{12}$$

The initial conditions for both cases are as follows.

$$q = 0 \text{ and } \frac{\partial q}{\partial t} = 0 \qquad \text{for } 0 < x < L,\ t = 0 \tag{13}$$

2.2. Calculation of the Laser Irradiance (Q_l)

When the laser light irradiation is absorbed within a very small depth of tissue ($\sim$1 μm), the laser heating can be predicated by considering the laser irradiation as a surface heat flux on the irradiated surface (see Eq. (9)). When the scattering is considerable over the visible and near infrared wavelength [14], the heat flux boundary condition is not enough to describe the laser deposition into a tissue. Rather, the laser light attenuation depends on the properties of laser light and its propagation. The absorbed laser is considered as a body heat source. The laser volumetric heat source can be determined as follows.

$$Q_L(x) = \mu_a \phi(x) \tag{14}$$

where μ_a is the absorption coefficient, and $\phi(x)$ is the local light irradiance varying with depth of the tissue.

To calculate light distribution in scattering tissue, a broad beam laser method [11] is adopted and the light distribution can be determined by the following relation.

$$\phi(x) = \phi_{in}[C_1 \exp(-k_1 z/\delta) - C_2 \exp(-k_2 z/\delta)] \tag{15}$$

where δ is the effective penetration depth; C_1, C_2, k_1, and k_2 are determined by Monte Carlo solutions, depending on the diffuse reflectance R_d; the effective penetration depth δ can be obtained from the diffusion theory as

$$\delta = \frac{1}{\sqrt{3\mu_a[\mu_a + \mu_s(1-g)]}} \tag{16}$$

where μ_s is the scattering coefficient and g is the scattering anisotropy. Equation (16) is valid when $0.1 \leq \mu_s/(\mu_a + \mu_s) \leq 0.999$ and $0.7 \leq g \leq 0.9$ [11].

2.3. Calculation of Thermal Damage Parameter (Ω)

The damage parameter is evaluated according to the Arrenius equation [1, 15].

$$\Omega = A \int_{t_0}^{t_f} \exp\left(-\frac{E}{RT}\right) dt \tag{17}$$

where A is the frequency factor, $3.1 \times 10^{98}\,\text{s}^{-1}$ [1]; E is the energy of activation of denaturation reaction, $6.28 \times 10^5\,\text{J/mol}$ [1]; R is the universal gas constant, $8.314\,\text{J/(mol} \cdot \text{K)}$; T is the absolute temperature of the tissue at the location where thermal damage is evaluated; t_0 is the time at onset of laser exposure; and t_f is the time of thermal damage evaluation. When $\Omega = 1.0$, the tissue is assumed irreversibly damaged which causes the denaturation of 63% of the molecules.

The following properties of a living biological tissue are used for this analysis. Thermophysical properties of tissues [17]: $\rho = 1000\,\text{kg/m}^3$, $k = 0.628\,\text{W/ (m K)}$, $c = 4187\,\text{J/(kg K)}$; thermophysical properties of the blood vessel: $\rho_b = 1060\,\text{kg/m}^3$, $c_b = 3860\,\text{J/(kg K)}$, $w_b = 1.87 \times 10^{-3}\,\text{m}^3/(\text{m}^3$ tissue s); optical properties [18]: $\mu_s = 120.0\,\text{cm}^{-1}$, $\mu_a = 0.4\,\text{cm}^{-1}$, $g = 0.9$; blood temperature: $T_b = 37°\text{C}$; and metabolic heat generation: $Q_m = 1.19 \times 10^3\,\text{W/m}^3$ [17]. The thickness of the slab of tissue is $L = 5\,\text{cm}$, and the initial temperature is $T_0 = 37°\text{C}$. The diffuse reflectance $R_d = 0.05$ is used for the laser light distribution of scattering tissue. Two laser irradiances are considered: $\varphi_{in} = 2\,\text{W/cm}^2$ and $30\,\text{W/cm}^2$. The laser duration time τ_L is 5 s. After the model convergence test, a total of 120 grid points and a time step (Δt) of 0.001 s are employed. Three different values of the coupling factor are taken based on the blood perfusion rate. According to the blood perfusion rate $w_b = 1.87 \times$ 10-3 $\text{m}^3/$ (m^3 tissue s), the values of ϵ are 0.0079, 0.025 and 0.0845 [12], and the coupling factors are 67,435, 55,078 and 47,488 $\text{W/m}^3\text{K}$ [12, 22].

The first case studied is that the laser light is highly absorbed by the tissue. As stated earlier, the heat flux boundary condition Eq. (9) is applied at the laser irradiated surface. Figure 2 compares the temperature responses at the irradiated surface obtained from the Fourier heat conduction, constitutive DPL model and generalized DPL model, and Figure 3 displays the change of the resulting thermal damage para-

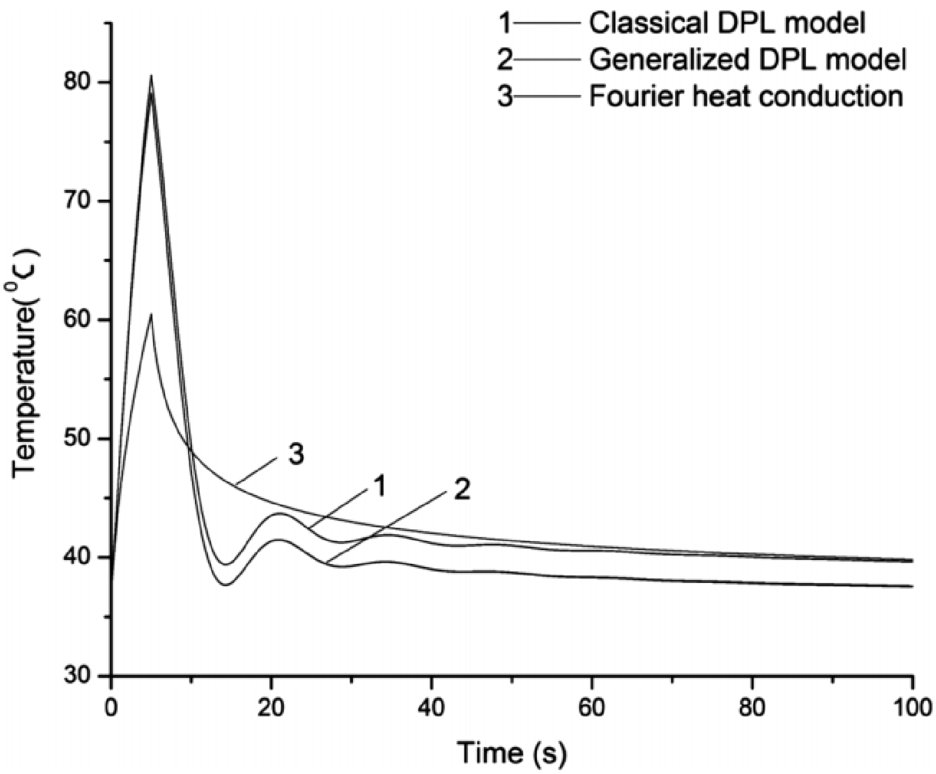

Figure 2. Temperature evolution at the irradiated surface of a highly absorbing tissue.