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OPTIMISATION OF LASER PARAMETERS FOR STUDYING THE TEMPERATURE PROFILES IN THE TISSUE

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ABSTRACT

In this study a model is proposed to estimate and optimise the exposure duration of laser beams of different powers for superficial Basal Cell Carcinoma (a skin tumour) of varying radii and depths. The frequent application of Laser in medical field makes it a matter of significance, to study the distribution of temperature in tissue and its effect on tumourous tissue during laser therapy. The tissue temperature is numerically estimated by solving Penne's bioheat equation for 2-D multi-layered skin model using Finite Volume Method. This study focuses on Basal Cell Carcinoma (BCC) which is a semi-malignant tumour of the uppermost layer of the skin i.e., epidermis. In this paper, different sizes of Basal Cell Carcinoma are considered which are irradiated by continuous wave Nd:YAG laser with Gaussian beam profile. The thermal damage to the tissue is estimated by using the Henriques' theory of skin burns. A grid-based model is employed to depict the temperature profiles against the tumour-tissue interface. An inverse proportional relation between laser power and irradiation time, for a particular tumour radius, was established in this study.

Keywords: Basal cell carcinoma, Bio-heat equation, Finite volume method, Laser, Thermal damage

I. INTRODUCTION

The interaction of laser with biological tissue is a subject of much interest owing to the ever increasing role of laser in therapy. Heat therapy for treatment of skin diseases has been in use for a long time. Studies have suggested that laser radiation can be used as a source of localised heat to produce localised hyperthermia that can be utilised to produce desired changes in the target tissue, thus making it an efficient therapeutic alternative [1–3]. Laser therapy is widely used as a general cutting tool in dermatology, gynecology, neurology, oncology [4, 5] and many other medical fields due to desirable qualities such as increased precision and improved haemostasis which ensures minimal healthy-tissue manipulation.

So, computational modelling becomes essential for understanding and predicting temperature evolution during laser therapy. Such an approach attempts to provide therapeutic models of increasing efficiency by optimising

laser parameters in accordance with tissue parameters like the size of the tumour, and its optical and thermal properties. Lasers are used in the treatments of the superficial tumours like basal cell carcinoma (BCC) [6]. Basal Cell Carcinoma (BCC) is a slow-growing form of skin cancer that starts from basal cells which are in the deepest layer of the epidermis. This kind of carcinoma consists of small islets of basaloid tumour cells which have well defined boundaries against the normal epithelia. For an efficient laser therapy of BCC, the laser parameters such as power, irradiation time, spot size, energy per pulse, repetition rate, and number of pulses are optimised based on the tissue optical characteristics such as anisotropy, scattering coefficient, absorption coefficient and refractive index.

The rate of thermal distribution is estimated based on Beer-Lambert's law [7, 8]. Out of the many available lasers which can be explored for thermal therapy, the Nd:YAG laser has been the most frequently used in medical applications [4, 5, 9] due to its potential to stop internal haemorrhage, to treat tumours, and to cause photo-coagulation. The Nd:YAG laser can produce energy at wavelength of 1064m*m*. When the tissue surface is exposed to laser, the tissue temperature starts to rise and phenomena like tissue coagulation, vaporisation, carbonisation, and burn occur sequentially resulting in the necrosis of tissue. In laser surgery, tissue coagulation is defined as the irreversible damage caused when temperature reaches above 60^{o}C upto 100^{o}C.

A literature survey was carried out to determine various models available and to compare their performances. A three step model for laser-tissue interactions was proposed by Welch [10] which included (1) modeling of heat deposition in tissue based on the absorption coefficient (μ_a) and scattering coefficient (μ_s), (2) determination of temperature response modeled by the heat conduction equation, and (3) estimation of tissue denaturation. Then a model was proposed by Takata [11] that predicted transient temperature and thermal damage. This model considered tissue modifications and steam blister formation. The results obtained from experiments conducted on pig skin using CO_2 laser validated the model. A model was developed by Halldorson and Langerholc [12] which considered modification of tissue properties after coagulation. Then a model for the study of thermal effects of Nd:YAG laser on biological tissues was presented by Cummins and Nauenberg [13]. This model differed from Halldorson and Langerholc in its discussion of the effect of modifications of tissue properties but it was validated using thermal data obtained by Halldorson and Langerholc on dog stomach [14]. A new model was developed by Jacques [15] who estimated the steady-state temperature in biological tissue with no damage prediction. The comparison between the temperature profiles of continuous wave (CW) and pulse laser was carried out in a work carried out by Banerjee et al. and it was found out that short pulse laser results in more localized heating than a continuous laser with a high peak temperature [16]. Finite Volume Method is used to monitor the heat flux distribution inside the two-dimensional rectangular medium which is absorbing and isotropically scattering by Muthukumaran et al. [17]. In a recent work the dual phase lag based heat conduction model alongwith the radiative transfer equation (RTE) to determine the propagation of light inside the tissue phantoms is developed by Kumar et al. [18]. Many of these models used complex mathematical solutions and some used simple solutions but did not discuss about the heat transfer for special cases of clinical applications like tumour treatment using thermal therapy.

The model developed here considers the interface at different penetration depths of the tumour inside the tissue layers and monitors the temperature front propagation with respect to the tumour boundary, depending on the exposure time for a particular laser power. The objective of this work is to develop a mathematical model to

estimate the temperature distribution inside the tissue during the thermal therapy of Basal Cell Carcinoma when irradiated with Continuous Wave (CW) Nd:YAG laser having Gaussian beam profile. The finite volume method has been used to solve an axisymmetry bio-heat equation for 2-D tissue slab to predict temperature.

II. MATHEMATICAL FORMULATION

1.1. Physical description and Model:

In this mathematical model, a slab of three-layered biological tissue having a BCC of a particular depth and spreading radius at the surface is irradiated with Gaussian laser beam profile. The tissue surface is assumed to be a perfectly insulated interface and the thermal energy distribution is supposed to be axisymmetric. The bio-heat equation for 2-D three-layer tissue model is solved using finite volume method. The implicit three time level method is used to solve the unsteady part to obtain the second order accuracy in time. When the laser beam is impinged on the tissue surface, the thermal response of a biological tissue is given by Pennes bio-heat equation as below.

$$\rho.\frac{\partial H}{\partial t} = k.\nabla^2 T + S - \dot{w}_b \qquad (2.1)$$

where ρ is the density of the tissue, H is the total enthalpy, T is the temperature, S is the source term appearing due to the laser irradiation, k is the thermal conductivity of tissue, c_p is the specific heat, $\dot{w}_b$ is the blood perfusion rate.

1.2. Source Term:

The source term comes into play because of laser-tissue optical interactions like reflection, absorption, and scattering. Many authors have developed mathematical models using different methods like Beer-Lambert's law, seven-flux model, Kubelka-Munk theory, photon diffusion approximation, multiple scattering, and Monte Carlo model [19]. The selection of model relies on the laser parameters like wavelength and the precision required as well as the optical and thermal properties of the tissue. The laser light experiences scattering when impinged on the biological tissue. There have been studies on animal skin in vivo [20], canine and human myocardium [21] to confirm the scattering of wavelengths in the range of IR and visible spectrum. The beam broadening model proposed by Yoon et al. [19] considered the scattering phenomenon along with the absorption of light, and beam broadening occurred due to multiple scatterings. The intensity (I) at the point (r,z) can be determined by the relation:

$$I(r,z) = (1-R) \times I_0.e^{-0.5(r/\tau(z))^2}.e^{-\gamma.z} \qquad (2.2)$$

where I(r,z) is the intensity or irradiance, R is the direct reflection coefficient from the tissue surface, g is the extinction coefficient and is equal to $\mu_a+\mu_s$, where ma and μ_s are absorption and scattering coefficients respectively, $\tau(z) = \tau_0.e^{(-\mu_s/2).z}$, τ_0 is the standard deviation of beam.

1.3. Thermal Parameters of dermis and sub-cutaneous fat:

The three important thermal parameters of tissue viz., density (r), specific heat (c), and thermal conductivity (k) are influenced by the water content of the tissue, the initial value of which is considered to be 70% [22]. The water starts to evaporate when temperature exceeds 100^oC, because of vaporisation. The model proposed by Dua and Chakraborty [23] described phase change processes, like melting and vaporisation, in treatments with high irradiance. In photo-thermal therapy, where temperature can exceed the vaporisation temperature of water

inside the tissue, photo-vaporisation can occur and hence the water fraction changes with time. These three key tissue thermal parameters for different water fractions at different time instances of thermal therapy can be described as [24]:

$$\rho\left(kg/m^3\right) = (6.16 \times 10^{-2}W + 0.938)^{-1} \quad \textbf{(2.3)}$$

$$c_p(J/kgK) = 2.5W + 1.7 \quad \textbf{(2.4)}$$

$$k(W/mK) = \rho \times 10^{-5}(0.454W + 0.174) \quad \textbf{(2.5)}$$

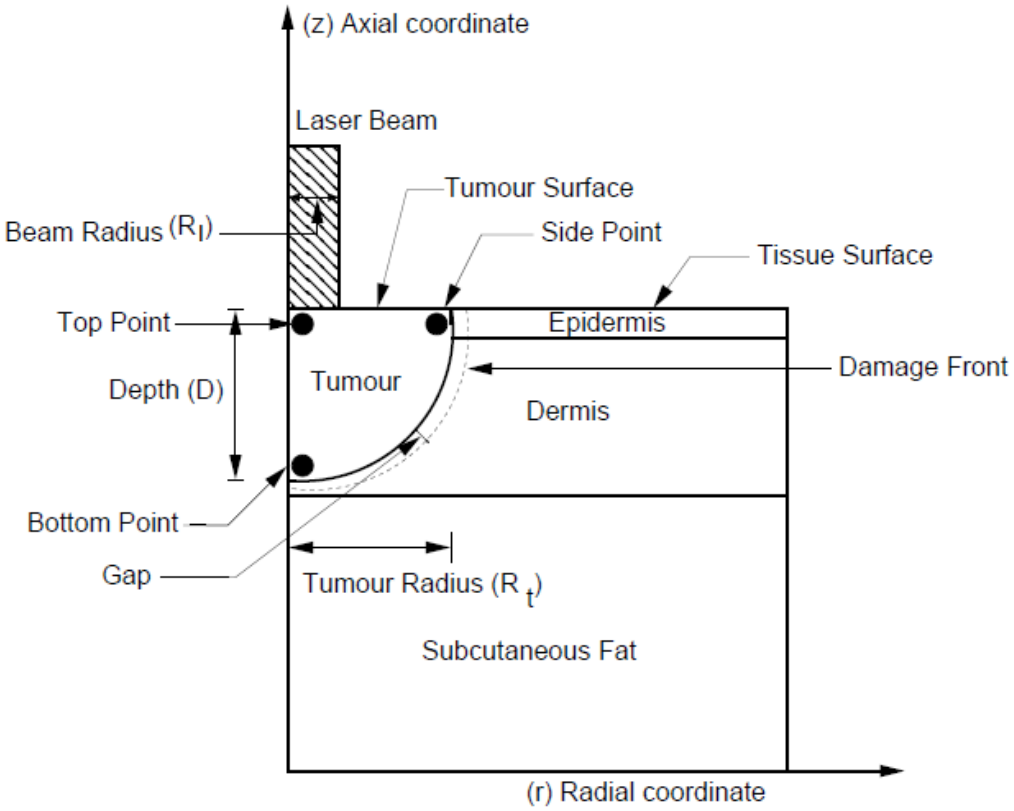


Figure 2.1: Schematic diagram of 2-D three layer skin model.

The "equation 2.1" is solved using finite volume method for 2-D biological tissue slab. There are many other analytical methods to solve this equation but their computational time hinders the determination of the temperature during the transient state. To minimise the computational time, numerical solutions which can approximate the exact solution and determine the transient temperatures in short time, can be used.

The cross-section of 2-dimensional three-layer skin model with the basal cell carcinoma penetrating inside the dermis, on which the laser beam is irradiated, is shown in "Fig. 2.1". The depth (D) is taken from the level of normal skin surface. The tumour depth (D) is considered equal to tumour radius (R_t) for this study. The beam radius (R_l) is kept 1.75 times the tumour radius, i.e. Beam Radius (R_l) = 1.75×Tumour Radius (R_t). The three black dots marked as top point, side point, and bottom point are shown for which the temperature is monitored throughout the laser exposure and even after the laser irradiation is terminated. The TABLE 1 shows the series of biological phenomena which occur during the photo-thermal therapy at different temperatures. For this case,

the temperature of the tissue reaches up to 100°C, so the biological effects considered are bio stimulation, hyperthermia, and reduction in enzyme activity, protein denaturation and vaporisation.

TABLE 1: Thermal effect of laser radiation [25].

Temperature (°C)	Biological Effect
37	Normal
<43	Biostimulation
43-45	Hyperthermia
50	Reduction in enzyme activity
60	Protein Denaturation (Coagulation)
70-80	Welding
80	Permeabilisation of Cell Membranes
100	Vaporisation
>150	Carbonisation
>300	Rapid cutting and Ablation

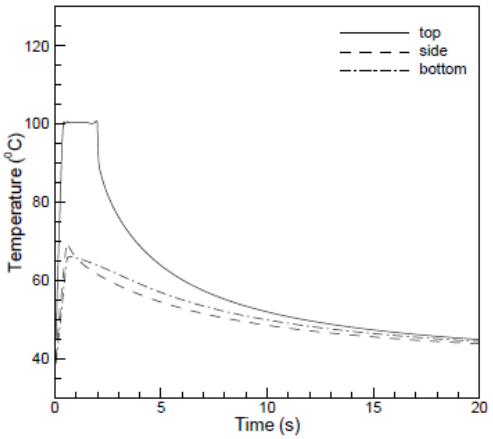
III. RESULTS AND DISCUSSION

The bio-heat equation is solved using finite volume method for 2-D axisymmetric three-layered Skin model with tumour of specific depth starting from the surface of the skin. The tissue optical properties are: $\mu_{a\cdot epi}$ = 571.29cm^{-1}, $\mu_{a\cdot derm}$ = 41.24cm^{-1}, $\mu_{a\cdot sub\ f\ at}$ = 10.0cm^{-1}, $\mu_{s\cdot derm}$ = 732.31cm^{-1} and $\mu_{s\cdot sub\ f\ at}$ = 918.58cm−1, and the laser parameters for CW Nd:YAG laser are: beam radii =2.625mm, 3.0625mm and 3.5mm for tumour radii = 1.5mm, 1.75mm and 2.0mm respectively. The average power is varied between 5.0−10.0W; the laser powers are taken at a step size of 1.0 for this study. The values of key thermal properties of tissue like density (r), specific heat (c_p), and thermal conductivity (k) depend on the water content which can be calculated using "equations 2.3, 2.4, and 2.5" respectively. The initial temperature of the tissue is taken as 37°C. The laser irradiation time for which the damage front moves 0.2mm beyond the tumour interface into the surrounding healthy tissue is considered to be the optimised irradiation time. This margin is taken into account so as to minimise the recurrence rate. In the plots where gap is plotted, the profiles are shown for the five time instances: (*i*) at the half of the irradiation time, (*ii*) at the time instance when laser is terminated, (*iii*) at the time instance where the damage front just touches the tumour interface near the q = 0°, (*iv*) at the time instance after which the damage front almost stops to propagate further and, (*v*) at time instance after which it totally stops further penetration.

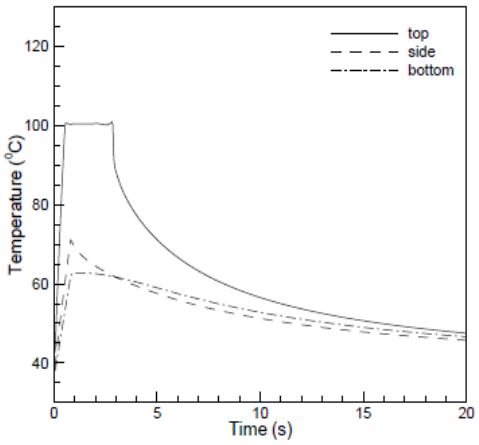
1.4. Temperature profiles with respect to time at top, side and bottom points near the tumour interface:

The variation of temperature with respect to time at three different locations, *viz.* top, side, and bottom points as shown in "Fig. 2.1", for three different tumour radii is depicted in "Fig. 3.1". The laser power for this case is 9W. At the point of interaction, temperature went up to 100°C and then it remained constant for some time, as the water inside the tissue started to vaporise and the temperature dropped drastically when the laser exposure is terminated. This is happening because the heat conduction into the tissue and heat lost into the air are counterbalanced by the heat supplied by the laser source. Once the laser exposure is terminated, two phenomena occurs *viz.*, the heat diffusion into the tissue and the heat dissipation into the air through surface due to which

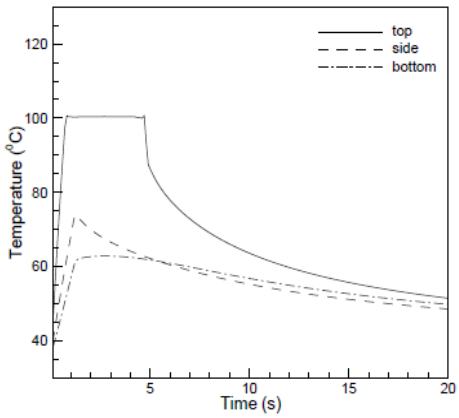
there is drastic decrease in the surface temperature. As the time progresses, after the laser termination, the temperature decreases at a slower rate because with the passage of time temperature gradient at the surface of the tissue decreases. As a consequence, the rate of heat diffusion decreases. The side and bottom points being at some distance from the point of interaction of laser beam with the tissue, the value of maximum temperature is around 70^{o}C and 65^{o}C for the side and bottom points respectively. It is also interesting to observe that the temperature reduction, after the termination of laser exposure, is rapid for side point as compared to bottom point. The reason for this is that, side point is close to the surface therefore, heat transfer will be governed by two phenomena:



a) Tumour radius = 1.5mm



b) Tumour radius = 1.75mm

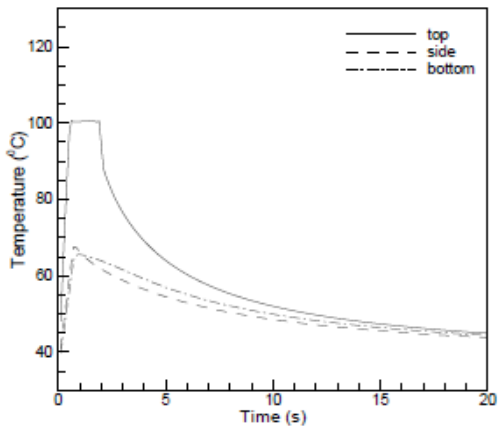


c) Tumour radius = 2.0mm

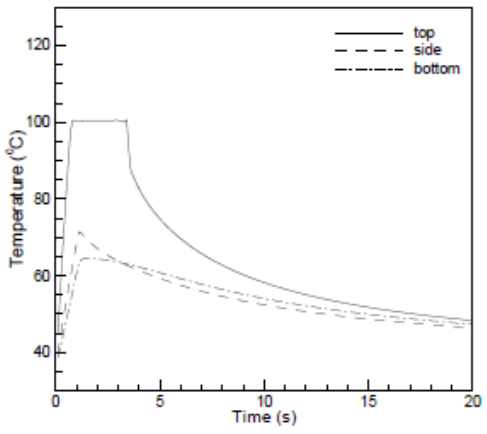
Figure 3.1: Temperature profiles for average Laser Power = 9W

heat diffusion into the surrounding tissue and heat dissipation into air. However, for the bottom point only one phenomenon, i.e. heat diffusion into the surrounding tissues, accounts for the heat loss, which is why the temperature decreases at a faster rate for side point as compared to the rate of temperature decrease for the bottom point. In "Fig 3.1(b)", it can be seen that the temperature profile stays at 100°C for a longer duration as compared to "Fig 3.1(a)". This stay is even longer in the profile of temperature plotted for 2.0mm tumour radius as shown in "Fig 3.1(c)". This is because of the larger irradiation time taken to achieve complete necrosis of the tumourous tissue with the increase in tumour size.

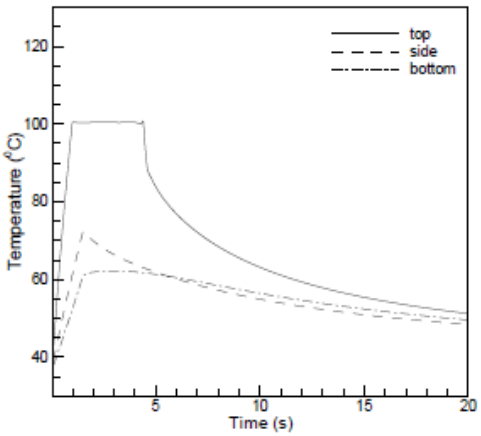
Furthermore, owing to the presence of unvaporised water in the tissue, there is no drastic rise in temperature. It can also be observed from these plots that the difference in the maximum temperatures achieved for side and bottom points is low for tumour radius of 1.5mm, and as the tumour radius increases, the maximum temperature difference between these two points also increases. This is due to the fact that the side point is experiencing a higher temperature because of the higher scattering coefficient of the tissue while, with the increase in the tumour radius, the distance between the bottom point and the interaction point increases which results in less temperature rise at this location.



a) Tumour radius = 1.5mm



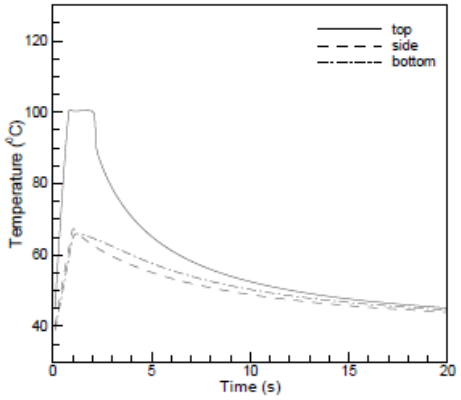
b) Tumour radius = 1.75mm



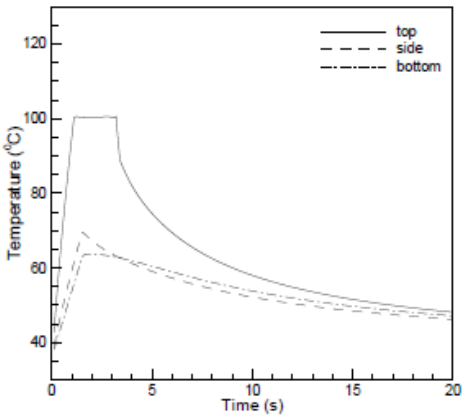
c) Tumour radius = 2.0mm

Figure 3.2: Temperature profiles for average Laser Power = 7W

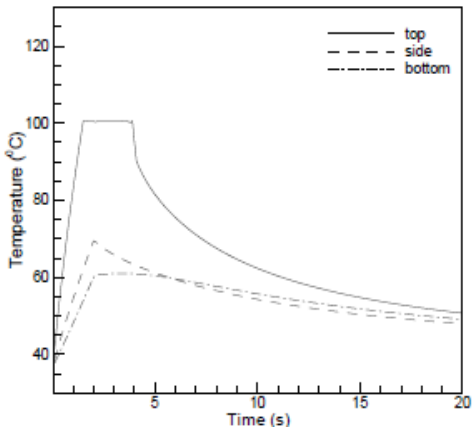
The temperature variations for the laser power of 7W are shown in "Fig. 3.2". In these plots it can be noticed that maximum temperature for the side point and bottom point for a tumour radius of 1.5mm is 68°C and 65°C; it is 73°C and 65°C for a tumour radius of 1.75mm, and 73°C and 60°C for a tumour radius of 2.0mm respectively. As noted earlier, the difference between the maximum temperatures at the two points is increasing with the increase in the tumour radius.



a) Tumour radius = 1.5mm



b) Tumour radius = 1.75mm



c) Tumour radius = 2.0mm

Figure 3.3: Temperature profiles for average Laser Power = 5W

Depicted in "Fig. 3.3" are the profiles of temperature for 5W laser power for different tumour radii. The maximum temperature for tumour radius of 1.5mm at side and bottom points is almost same and is equal to 66°C. The maximum temperature achieved for these two points for 1.75mm tumour radius is 70°C and 64°C respectively; for tumour radius 2.0mm, it is 70°C and 60°C respectively. Also, it is illustrated in these profiles that the rate of temperature rise for all the three points decreases as the tumour radius is increased for a particular power because energy per unit area per unit time decreases.

The laser irradiation time for which efficient tumour necrosis is obtained using laser powers between 5.0W −10.0W for three different tumour radii is listed in "TABLE 2". The laser powers are taken at a step size of 1.0 for this study. It can be noticed that as the laser power increases for a given radius, the irradiation time decreases almost linearly.

TABLE 2: Optimised exposure duration for lasers of different average powers for various tumour radii

Tumour Radius (mm)	Average Power (Watts)					
	5	**6**	**7**	**8**	**9**	**10**
1.5	1.0	0.90	0.70	0.65	0.55	0.50
1.75	1.60	1.20	1.10	0.90	0.85	0.80
2.0	2.10	1.80	1.60	1.40	1.20	1.0

IV. CONCLUSION

A mathematical model that predicts the temperature and monitors the propagation of thermal damage front during the thermal therapy for a particular tissue has been developed. This model reflects the understanding of thermal events like coagulation and vaporisation when the laser of a specific power is irradiated on a tumourous tissue. In this study, the optimised irradiation time has been predicted for a particular tumour radius, laser beam radius, laser power and specific skin type. It can be concluded from the results that irradiation time decreases almost linearly with increase in the laser power for a given tumour radius. The laser beam radius also affects the damage extent and the beam radius has to be optimised for a particular tumour radius so that damage front propagates uniformly. The present study also proposes a correlation between optimised exposure time and laser

power for a particular tumour radius, which is useful to the dermatologist in deciding the proper process parameters for an efficient treatment of basal cell carcinoma.

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