A new Monte Carlo code for absorption simulation of laser-skin tissue interaction

Afshan Shirkavand[1], Saeed Sarkar[1,2], Marjaneh Hejazi[1], Leila Ataie-Fashtami[3,4], and Mohammad Reza Alinaghizadeh[5]

[1]*Medical Physics & Biomedical Engineering Department, Tehran University of Medical Sciences, Tehran, Iran*

[2]*Research Center for Science and Technology in Medicine, Tehran, Iran*

[3]*Iranian Academic Center for Education, Culture and Research, Tehran, Iran*

[4]*Iranian Center for Medical Laser, Tehran, Iran*

[5]*NOOR Medical Imaging Center, Tehran, Iran*

Received October 9, 2006

In laser clinical applications, the process of photon absorption and thermal energy diffusion in the target tissue and its surrounding tissue during laser irradiation are crucial. Such information allows the selection of proper operating parameters such as laser power, and exposure time for optimal therapeutic. The Monte Carlo method is a useful tool for studying laser-tissue interaction and simulation of energy absorption in tissue during laser irradiation. We use the principles of this technique and write a new code with MATLAB 6.5, and then validate it against Monte Carlo multi layer (MCML) code. The new code is proved to be with good accuracy. It can be used to calculate the total power absorbed in the region of interest. This can be combined for heat modelling with other computerized programs.

OCIS codes: 170.0170, 170.3660, 170.5280, 170.1610.

Since Wilson and Adam first introduced Monte Carlo simulations into the field of laser-tissue interaction, it has been widely used to simulate light transport in tissues for various applications and gone through several improvements[1−3]. Because of the flexibility of the Monte Carlo algorithm to handle various physical conditions, it is chosen to simulate photon distribution inside the medium[4].

Laser as a tool is becoming available to a growing number of physicians, but before the doctor can use this tool he must select a laser, a beam power, a spot size, and an irradiation time. Since small differences in any of these parameters can determine whether an application is disastrous, some *a priori* knowledge about the effects of each parameter is needed. This information is usually provided by mathematical models. Any model of a laser treatment is based on the distribution of light in the tissue[4].

This paper describes the Monte Carlo method for modelling absorbed light in skin tissue. The formulas necessary for implementation of the Monte Carlo method in computer code are provided. The structure of the Monte Carlo code written by MATLAB6.5 and the results for validating the Monte Carlo implementation are given. The goal of this study is to simulate the absorption of hair removal diode laser in skin tissue.

The algorithm is mainly based on the technique proposed by Jacques *et al.*[2,3]. The rules used in Monte Carlo programs for photon migration in biological tissue are straightforward as described in Ref. [3]. Differences between programs arise in how the photon fluence or flux is recorded during the Monte Carlo simulation. Below, we briefly describe the rules for photon migration and how we record the photon fluence within the medium and the flux of photons exiting the medium. A two-dimensional model to simulate the propagation of light in participating media is developed. The information provides physical details about the photon absorption in skin tissue during the laser irradiation[5].

The flow chart of a Monte Carlo program that has been written in MATLAB language is shown in Fig. 1. Once launched, the photon is moved a distance Δs where it may be scattered, absorbed, propagated undisturbedly, internally reflected, or transmitted out of the tissue. The photon is repeatedly moved until it either escapes from or is absorbed by the tissue. This process is repeated until the desired number of photons has been propagated.

Different steps used in this program are defined briefly below.

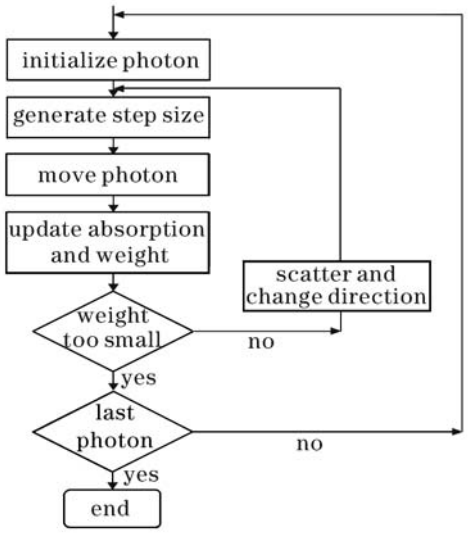

Fig. 1. Flow chart of the variable step size Monte Carlo technique.

1671-7694/2007/040238-03

Photon initialization. The Monte Carlo method begins by launching a photon into the tissue. If a collimated beam normally incident on a slab is simulated, then the photon's initial direction is chosen downwards into the tissue. One simple technique to improve the efficiency of a Monte Carlo program is to propagate many photons (a packet) along each pathway. Usually only one photon follows each pathway, and at each step the photon may be either absorbed or scattered. If a photon packet follows each pathway, then some portion of the packet will be absorbed at each step. The size of this packet is called the weight (w) of the photon. Its initial weight is set to unity[2], i.e. $1 = w$. The effect of a refractive-index-mismatched interface between the medium and the ambient is neglected in this study. However, specular reflection due to mismatched boundary can be easily taken into account. The portion of the laser beam will be reflected at the boundary. Thus the remaining photon weight to propagate through the medium is[2]

$$w = 1 - R_{\rm sp}, \tag{1}$$

where $R_{\rm sp}$ is the specular reflectance due to the different refractive index between the air (n_1) and turbid medium (n_2)[2]

$$R_{\rm sp} = \frac{(n_1 - n_2)^2}{(n_1 + n_2)^2}. \tag{2}$$

Generating step size. A variable step size method was used for writing the code. In this method, a different step size for each photon step is chosen. The probability density function for the step size follows Beer's law (i.e., it is more likely for a photon to travel a short distance than a long distance), and the probability is proportional to an $e^{-\mu_t \cdot \Delta s}$ function of a random variable (ζ) uniformly distributed between zero and one. Thus, a random variable with this distribution is yielded as

$$\Delta s = \frac{-\ln \zeta}{\mu_t}, \tag{3}$$

where the step size Δs represents the distance that a photon will travel before interacting (through absorption or scattering) with the tissue.

Photon moving. It is convenient to describe the photon's spatial position with three Cartesian coordinates and the direction of travelling with three direction cosines. The required formulas for propagation are simple, and the angle variables describing photon direction do not change unless the photon's direction changes. The direction cosines are specified by taking the cosine of the angle that the photon's direction makes with each axis. These are specified by μ_x, μ_y and μ_z corresponding to each of the x, y and z axes, respectively. For a photon located at (x, y, z) travelling a distance Δs in the direction (μ_x, μ_y, μ_z), the new coordinates (x', y', z') are given by

$$x' = x + \mu_x \Delta s, \quad y' = y + \mu_y \Delta s, \quad z' = z + \mu_z \Delta s. \tag{4}$$

Photon absorption. It was assumed that at each photon interaction in each step size, a fraction of its energy becomes absorbed and the rest scattered. The fraction of the packet that is absorbed is

$$\Delta w = \frac{\mu_{\rm a}}{\mu_{\rm a} + \mu_{\rm s}}. \tag{5}$$

The new photon weight w' is given by $w' = w - \Delta w$, which represents the fraction of the packet that is scattered on this step.

An absorption event requires that both the location and the amount of light absorbed be recorded. In this technique we saved each absorbed fraction of photon in each interaction in the absorption matrix that its dimensions were equal to the spatial resolution of target matrix.

Photon termination. After updating absorption of photon, its weight is compared with a threshold. If the weight of the photon is below the threshold, and if it is the last photon, the program will be ended. But if it is not the last one, this process becomes repeated.

Scattering of the photon. A normalized phase function describes the probability density function for the azimuthal and longitudinal angles for a photon when it is scattered. If the phase function has no azimuthal dependence, then the azimuthal angle is uniformly distributed between 0 and 2π, and may be generated by multiplying a pseudo-random number uniformly distributed over the interval zero to one by 2π ($\phi = 2\pi\zeta$). The azimuthal angle μ for an isotropic distribution is given by[1−3]

$$\cos\theta = 2\zeta - 1. \tag{6}$$

Since scattering in tissue is characterized by the Henyey-Greenstein phase function, the generating function is[1−3]

$$\cos\theta = \frac{1}{2g}\left\{1 + g^2 - \left[\frac{1 - g^2}{1 - g + 2g\zeta}\right]^2\right\}, \tag{7}$$

where g is the anisotropy factor.

If a photon is scattered at an angle (θ, ϕ) from the direction (μ_x, μ_y, μ_z) in which it is travelling, and assuming that the angle is too close to the normal, then the new direction (μ'_x, μ'_y, μ'_z) is specified by

$$\mu'_x = \sin\theta\cos\phi, \quad \mu'_y = \sin\theta\sin\phi, \quad \mu'_z = \frac{\mu_z}{|\mu_z|}\cos\phi. \tag{8}$$

Once the Monte Carlo simulation is complete for all the photons, an absorption power density matrix is generated for a given tissue configuration. This matrix represents the amount of laser power absorbed by the region in the form of power density.

In a cylindrical system, the absorption power density matrix is a function of the coordinates r and z, with a cylindrical symmetry. Multiplying the absorption photon probability density by the desired laser power, the total power absorbed in the region is obtained[6].

After the writing process of code was completed, to examine the validation of this code we simulated the published modelling[7] with our code. In fact, we studied their methods step by step and then extracted any crucial details. Then we set simulations based on the information. Klavuhn *et al.* developed a theoretical analysis using Monte Carlo model and solving the heat transfer equation to construct a more complete role of sapphire cooling disk during laser treatment of hair removal[7]. They used a three-layer model of skin tissue consisting of 0.080-mm-thick epidermis, 0.020-mm-thick basal layer, and 4.9-mm-thick dermis. They assumed a cylindrical

Table 1. Optical Properties of Skin Used in Multi-Layered Skin Tissue Modelling at the Wavelength of 800 nm in Ref. [7]

	μ_a (cm^{-1})	μ_s (cm^{-1})	g	n
Epidermis	0.3	40	0.8	1.37
Basal Layer	3	40	0.8	1.37
Dermis	0.3	40	0.8	1.37

hair shaft with 0.200-mm diameter at the center part of the model. Optical properties used in this model are summarized in Table 1. They modeled laser beam as a square pulse of constant irradiance $E_0 = 2000$ W/cm^2 over the pulse duration of 30 ms that results in a total fluence of 60 J/cm^2. They used a non-uniform 30×60 grid representing the physical domain to effectively resolve large temperature gradients. The program was run on a computer for 3×10^6 photons.

Figure 2 shows the result of validating the code by giving the fluence rate distributions in skin depth. Our calculation was done on the millimeter scale. The result using standard Monte Carlo multi-layer (MCML) method is also shown in the figure for comparison. According to the figure, the fluence rate within the skin drops rapidly with depth. Although the decrease of the fluence rate is more considerable after the depth of 0.2 mm, the rapid increase of the fluence rate is approximately equal for two codes. They have both shown a maximum fluence rate at the depth of basal layer of which the absorption coefficient is the greatest one. According to the t-test results, the simulation results related to the depth below 1 mm are evaluated with $\alpha < 0.05$, showing no significant difference with standard method MCML. But the results of other depths have significant difference with MCML.

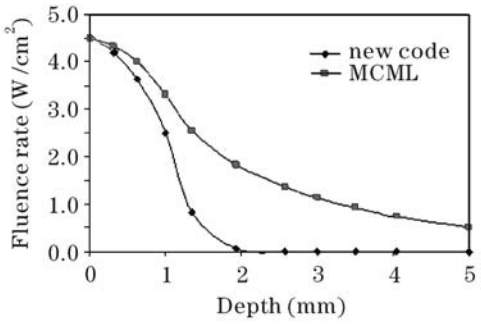

Fig. 2. Calculated fluence rate distributions of 800-nm radiation along the center line of a 10-mm diameter beam in tissue.

For the theoretical analysis presented herein, there are some assumptions and limitations. For example, the skin surface was assumed to be smooth, and any back reflection from within the sapphire was neglected. In practice, however, some of the photons scattered out of the skin would be reflected back into the skin (i.e., recycled) and the actual coupling of photons into and out of the skin would depend somewhat on the roughness of the skin surface.

It is presumed that this new code can be appropriate for modelling the absorption and flux distribution in skin tissue. It also can be combined with the finite difference method for solving the heat diffusion equation to obtain the temperature distributions in target tissue and surrounding tissue during laser photo-thermal interaction. It also has a benefit to use this code for different target properties and also different laser irradiation parameters.

The authors would like to express their sincere gratitude to Mr. Sarkarati who was a great help in developing this study. A. Shirkavand's e-mail address is ashirkavand @razi.tums.ac.ir or afshanshirkavand@gmail.com.

References

1. B. C. Wilson and G. Adam, Medical Physics **10,** 824 (1983).
2. S. A. Prahl, M. Keijzer, S. L. Jacques, and A. J. Welch, Proc. SPIE **5,** 102 (1989).
3. S. L. Jacques and L. Wang, in *Optical-Thermal Response of Laser-Irradiated Tissue* A. J. Welch and M. J. C. van Gemert, (eds.) (Plenum, New York, 1995).
4. S. A. Prahl, *"Light Transport in Tissue"* PhD Thesis (University of Texas at Austin, 1988) pp.1—210.
5. D. A. Boas, J. P. Culver, J. J. Stott, and A. K. Dunn, Opt. Express **10,** 159 (2002).
6. J. J. Crochet, S. C. Gnyawali, Y. Chen, E. C. Lemley, L. V. Wang, and W. R. Chen, J. Biomed. Opt. **11,** 034031 (2006).
7. K. G. Klavahn and D. Green, Laser in Surgery and Medicine **31,** 97 (2002).