Diagnosing different types of skin carcinoma based on their optical properties: A Monte-Carlo implementation

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Abstract. Skin cancer is a very common and serious type of cancers worldwide. Among many kinds of non-melanoma skin cancers, Basel Cell and Squamous Cell Carcinoma are highly treatable in case of early detection. Various Diagnosing techniques are employed to detect skin cancer, such as dermoscopy, OCT, biopsy and physical examination according to the medical case. However, the non-invasive optical methods are gaining validity due to their competitive advantages including safety and functionality. In addition, they are painless and high sensitive to the examined tissue metabolic changes. The propagation of light in any biological tissue is controlled by its optical absorption and scattering properties that highly depend on the wavelength of the utilized light. Monte-Carlo simulation is a forward numerical method used to describe light propagation in biological tissues depending on their optical parameters. In this work, Monte-Carlo simulation method was implemented to characterize the light propagation in normal dermis, Infiltrative Basal Cell Carcinoma, Nodular Basal Cell Carcinoma, and Squamous Cell Carcinomas in order to differentiate healthy from cancerous tissues. The obtained results provided information about the amount of light reflectance, transmittance, absorbed fraction and fluence rate distribution in the examined tissues showing different values at each condition over a wide range of wavelengths, which provide a simple, safe and functional tool for diagnosing these categories of skin carcinoma.

1. Introduction

Among different diseases affecting the human skin, skin cancer is enrolled as the most dangerous diseases. It mainly occurs because of the uncontrolled cell growth on the skin layers. The most problem of these abnormal cells is their ability to move or extent to other tissues and body organs [1,2]. Basal Cell Carcinoma “BCC” and Squamous Cell Carcinoma “SCC” are very common types of skin cancer, BCC commonly appears in the face but can also be found in neck, ears, head and shoulders and rarely causes a metastasis [3,4], however, it is not very dangerous if it is early detected. While, SCC occurs in the cells that form the upper layer of the skin and can extend to deeper layers. It mostly appears in body parts that is exposed to UV light such as legs or feet. If not early detected, SCC can reach a very large size and may cause metastasis [4].

In the field of medical diagnosis, the physician can determine the type of skin cancer through many techniques such as optical coherence tomography “OCT”, reflectance confocal microscopy, biopsy and dermoscopy [5]. High frequency ultrasound and Doppler sonography are non-invasive techniques implemented to characterize the skin according to the backscattered echo resulting from pulsed ultrasound [6]. While, in biopsy, a skin sample from the affected part is detached from the body to be
examined [6]. OCT is also a noninvasive imaging technique utilized to show the linear characteristics of the skin such as scattering, absorption birefringence and refractive index with acceptable resolution images [7,8]. As a minimally invasive technique, Raman spectroscopy is considered a powerful tool for clinical diagnosis of skin malignancies and in vivo skin cancer screening that almost doesn’t require sample preparation [8,9].

In addition to the conventional practical methods, Monte-Carlo (MC) simulation is a numerical method utilized to validate the theoretical basis of many skin cancer detection techniques [10–13]. MC method can simulate the diffuse propagation of light in single and/or multilayer tissue sample based on its optical absorption and scattering characteristics at a specific wavelength.

In this work, the Monte-Carlo simulation method was utilized to distinguish between the normal and diseased tissues. The differentiation criteria depended on the variations in optical absorption and scattering properties between healthy and cancerous skin tissues at laser wavelengths starting from 470 nm to 970 nm. The obtained reflectance, transmittance, absorbed fraction profiles of Infiltrative Basal Cell Carcinoma, Nodular Basal Cell Carcinoma, and Squamous Cell Carcinomas provided a promising diagnosing method for these types of skin carcinoma.

2. Methods

2.1. Monte-Carlo Implementation

Monte Carlo simulation for photon propagation in biological tissue is a statistical method based on the tracing the propagation of a large bundle of photons generated by a computer. It stimulates photon propagation in a highly scattering medium such as biological tissue by applying photon propagation rules including probability distribution and angle of deflection [14,15].

Photons on Monte Carlo stimulation are not treated by wave phenomena; however, the phase and polarization are ignored. The stimulation process starts with photon weight equals to unity, then, after the photon hits the surface of the tissue, a portion of photon will reflect and escape while the rest will internally reflect or scatter and propagate continuously. This process is repeated until the weight of photon become less than a specified threshold. The number of photons used in the stimulation is a matter of desired precision and the required spatial resolution. To implement Monte-Carlo, main optical information of the examined tissue must be known including, absorption coefficient, scattering coefficient, anisotropy, refractive index and the thickness of the sample[16].

In the present study, common types of skin cancers with previously published optical absorption and scattering parameters [17] have been examined. The selected tissue types are; Infiltrative Basal Cell Carcinoma, Nodular Basal Cell Carcinoma, and Squamous Cell Carcinoma with a sample thickness equals to 1 mm. Number of photons in MC implementation was 50000 photons.

2.2. Tissue optical parameters

Tissue absorption and scattering parameters can be determined from diffuse light measurements including reflection and transmission. These measurements are collected using either integrating spheres [18,19] or distant detectors [20–22] based techniques. Diffuse reflection and transmission are then utilized in suitable mathematical model to reconstruct the optical parameters. Inverse Monte-Carlo [23], Kubellka-Munk [24]and inverse adding doubling [25] are the most common mathematical models used for this purpose.

The optical parameters of many normal and diseased biological tissue types including human skin have been calculated over a wide range of wavelengths in [26]. Some of these published values have been introduced to our MCML implementation to differentiate between Infiltrative Basal Cell Carcinoma, Nodular Basal Cell Carcinoma, Squamous Cell Carcinoma and normal dermis.

2.3. Determining the Fluence rate distribution
Recent advances in describing laser transfer energy in biological tissues are based on transport theory of light propagation that utilizes the radiative transport equation RTE to represent this propagation [27]. The radiative transport equation can be written as. [28]

\[
s \nabla L(r,s) = - (\mu_a + \mu_s) L(r,s) + \mu_s \int p(s,s') d\omega' \frac{1}{4\pi}
\]

(1)

where \( L(r,s) \) is the radiance of light at position \( r \) traveling in a direction of the unit vector \( s \) and \( d\omega' \) is the differential solid angle in the direction \( s' \). \( \mu_a \) is the tissue absorption coefficient and \( \mu_s \) is the scattering coefficient.

When a laser beam is incident on a turbid medium as biological tissue, the radiance can be divided into a coherent and diffuse term. The diffuse radiance is the main problem which transport theory has to deal with, since scattered photons do not follow a determined path. Therefore, many approximations and statistical approaches were introduced to the transport equation depending on the value of the albedo of the medium [29].

Using radiative equation to calculate the light distribution in tissue requires the knowledge of the absorption and scattering coefficients as well as the phase function. In order to estimate these parameters, one must have a solution of the radiative transport equation. The diffusion equation (2) is an approximation to the radiative transport equation in which the radiance is assumed to be nearly isotropic, it is a partial differential equation that is significantly easier to be solved than the radiative transport equation [30].

\[
\frac{\partial \psi(\vec{r},t)}{\partial t} + \mu_s \psi(\vec{r},t) - \nabla \cdot [D \nabla \psi(\vec{r},t)] = S(\vec{r},t)
\]

(2)

where the constant \( D = \frac{1}{3(\mu_s + \mu_s')} \) is the diffusion coefficient and \( \mu_s' = (1 - g) \mu_s \) is the reduced scattering coefficient and \( g \) is the anisotropy factor. \( S(\vec{r},t) \) represents the source term and \( \psi(\vec{r},t) \) is the fluence rate.

In this study, the diffusion equation was solved using the finite element method [31,32] under the environment of COMSOL Multiphysics 5.2 software [33]. Helmholtz Equation (3) in COMSOL is employed to model the diffusion equation presented in (3) in the steady state.

\[
\nabla (-C \nabla u) + au = f
\]

(3)

Comparing the parameters in equation (3) with those in (2) we get, \( u = \phi, C = D, a = \mu_s \), and \( f = S \). A 1 cm circular model was created with a point source at the center of the model as presented in figure 1(a). The implemented mesh (figure 1(b)) has element size ranges from 0.000057 to 0.02 cm. Based on that predefined finite element model; the fluence rate (W/cm²) distribution at the surface of the various selected tissues was obtained.

\[\text{Figure 1. The implemented finite element (a) model, (b) mesh.}\]
3. Results and discussion

Using Monte-Carlo simulation, the spatially resolved steady state diffuse reflectance profiles for normal dermis, infiltrative, nodular, and squamous cell carcinoma were obtained at 470, 570, 670, 770, 870, 970 nm laser irradiation of 2.7 mm spot size as presented in figure 2. As illustrated from the figure, Normal dermis shows almost maximum reflectance values over the considered wavelengths. However, at 470 and 570 nm, the maximum reflectance was noticed in infiltrative basal cell carcinoma. Moreover, the lowest reflectance was observed in squamous cell carcinoma over the all wavelengths. Reflectance of nodular basal cell carcinoma always intermediates squamous cell carcinoma and infiltrative basal cell carcinoma.

![Diffuse reflectance profile of normal dermis, infiltrative, nodular, and squamous basal cell carcinomas at (a) 470 nm, (b) 570 nm, (c) 670 nm, (d) 770 nm, (e) 870 nm, (f) 970 nm.](image)

Figure 2. Diffuse reflectance profile of normal dermis, infiltrative, nodular, and squamous basal cell carcinomas at (a) 470 nm, (b) 570 nm, (c) 670 nm, (d) 770 nm, (e) 870 nm, (f) 970 nm.
The transmission and absorbed fraction in each studied case at selected wavelengths were also determined. Figure 3(a) and (b) illustrates the transmittance and absorption profiles for normal dermis, infiltrative, nodular, and squamous cell carcinomas respectively.

As conducted from figure 3, infiltrative basal cell, nodular basal cell, and squamous cell carcinoma almost take the same regime over the studied wavelengths. Unlike the normal dermis which takes the opposite behaviour from 750 to 950 nm for both transmission and absorption profiles. Moreover, infiltrative basal cell carcinoma has almost the maximum absorbed fraction and minimum transmission rather than other types of carcinoma. As more investigations are required for accurate differentiation, the fluence rate distribution at the surface of the examined tissues is obtained as presented in figure 4.
Figure 4. Fluence rate distribution at the examined tissue surface (a-d) at 470 nm, (e-h) at 570 nm, (i-l) at 670, (m-p) at 770, (q-t) at 870, (u-x) at 970
According to the optical fluence rate images of figure 4 and the curves presented in figure 5, squamous cell carcinoma gives the minimum fluence rate along the all used wavelengths followed by nodular basal cell. However, infiltrative basal cell almost gives the same fluence rate as the normal dermis. Therefore, these results can help in differentiating squamous cell and nodular basal cell carcinoma from the normal skin.

![Figure 5](image)

**Figure 5.** Maximum Fluence rate at different laser wavelengths

The obtained variations in diffuse reflectance profiles, transmittance, absorbed fraction and fluence rate distribution between the different tissue types are mainly due to the difference in the tissue’s absorption and scattering properties. These properties are wavelength dependent and highly related to tissue chromophores concentrations including haemoglobin, melanin, lipids and water contents [34]. Tissue diffuse reflectance is considered as a fingerprint for each tissue type and condition as it changes upon appearing any abnormality in the tissue [21]. Therefore, both tissue optical parameters and diffuse reflectance profile can powerfully assist in the process of medical diagnosis and therapy monitoring [35–38]. Therefore, our proposed results can offer additional diagnostic information compared to other traditional techniques.

### 4. Conclusion

Three types of skin carcinoma infiltrative, nodular and squamous basal cell have been compared with normal dermis using Monte Carlo stimulation for light propagation in biological tissue at different laser wavelengths. This study showed maximum reflectance for normal dermis at all employed wavelengths except for 470 nm and 570 nm, the maximum reflectance was for infiltrative basal cell carcinoma which can be used to differentiate between normal and cancerous skin. Moreover, light transmittance and absorbance have been studied. Absorbance and transmittance showed significant values variation for normal rather than cancerous tissues especially at wavelength range 750 nm to 950 nm out of this range values are very close.

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