In Vivo Measurement and Modeling of Multispectral Reflectance Images for Melanoma Diagnosis

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Abstract — Noninvasive detection of malignant melanoma in early stages is critical to improve patients’ prognosis. We acquired in vivo reflectance images of dysplastic lesions from 12 patients at 31 wavelengths from 500 to 950 nm. Based on these image data, we developed a parallel Monte Carlo code to simulate reflectance images from a heterogeneous skin tissue model. With this tool, we have investigated the dependence of the lesion contrast in the reflectance image on the heterogeneous distribution of tissue optical parameters. The Monte Carlo model is currently used to generate multispectral reflectance imaging data for multivariate analysis of the in vivo imaging data.

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It has been estimated that Caucasians may develop up to 50 clinically benign nevi by age 40. Patients with more than 100 nevi were estimated to have a 3-fold to 10-fold increased risk of developing malignant melanomas and pigmented basal cell carcinomas in comparison to the general population\textsuperscript{[1]}. Diagnosis of MM is currently established by histopathology of biopsied tissues from the suspicious-appearing nevi or pigmented lesions. These patients often present a difficult dilemma to primary-care physicians and dermatologists. A physician has to either prescribe painful and costly excision biopsy with likely cosmetic disfigurement with limited information on the lesion or leave untouched with the risk of MM developing in the patients. Therefore, cost-effective pre-biopsy methods of examination could greatly improve patient care and reduce medical cost with better specificity and sensitivity than what are available now. In this report, we present multispectral reflectance image data acquired from 12 patients with dysplastic lesions at 31 wavelengths from 500 to 950 nm and results of numerical studies of reflectance imaging method by a Monte Carlo (MC) code.

A multispectral imaging system employing a thermoelectrically cooled CCD camera has been constructed to acquire polarimetric images\textsuperscript{[2]}. Fig. 1 presents a schematic of the imaging system. We used a xenon fiber optic light source and a collimating lens to produce a parallel light beam of

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{schematic.png}
\caption{A schematic of the polarimetric multispectral imaging system. S: light source; L1, L2: lenses, F: wavelength filter; P: polarizer; h: incident angle; C: contact window.}
\end{figure}
25 mm in diameter with wavelengths ranging from 300 to 1100 nm for full-field illumination at the skin surface. The illumination beam was first passed through a long-pass filter with the edge at 500 nm and then one of two liquid-crystal-tunable-filters for selecting a waveband of about 10 nm bandwidth between 500 to 950 nm with a stepsize of 15 nm. The illumination arm of the imaging system was at an oblique angle of $\eta = 30^\circ$ from the normal of the skin surface while the imaging arm was oriented along the normal direction to reduce contribution to the image signals by skin surface reflection. Polarimetric images have been acquired from four patients suspected to have dysplastic nevi or MM and prescribed to be removed by their surgical oncologist. All participating patients agreed to be imaged for our study and were required to read and sign an informed consent form with the detailed description of the study before the start of imaging process. The patient study protocol and consent form were approved by the IRB of East Carolina University and followed strictly throughout this study. For each patient, a total of 64 images were acquired at 31 wavelengths from each patient between 500 and 950 nm with two polarimetric images, $I_{\perp}(x, y, \lambda)$ and $I_{//}(x, y, \lambda)$, at each wavelength. Immediately after patient imaging, the same sequence of multispectral images of a certified 40% diffuse reflectance standard was acquired to determine the relative distribution of the incident light at the object plane with the imaging polarizer set parallel to the incidence plane. With these images and the output power of the incident beam measured at each wavelength, we obtained the irradiance distribution of the incident light $I_i(x, y, \lambda)$. Examples of the multispectral reflectance image data from one patient at selected wavelengths is shown in Fig. 2.

![Reflectance images of $R_{//}$ of lesion #4 at 16 selected wavelengths noted at the lower right corner. Bar = 10 mm.](image)

Based on the in vivo imaging data, we have developed a parallel Monte Carlo code that can generate reflectance images by calculation of photon density distribution at the air side of a two-layer heterogeneous skin model, as shown in Fig. 3. The details of the MC algorithm and the skin model have been reported elsewhere [3–5]. In tracking photon in the turbid media of skin tissues, we used the Henyey-Greenstein function as the scattering phase function which is characterized by an anisotropy factor $g$. A collimated beam of diameter $2w = 25$ mm is incident at a angle of $\eta = 30^\circ$ to the surface normal. In the above defined heterogeneous skin model, we designed a central region of cylindrical shapes of radius $r$ with optical parameters different from the peripheral region in the illuminated area to imitate a pigmented lesion in a normal tissue. The images in Fig. 4 exhibit clear differences in reflectance $R$ between these two regions in which stronger light absorption in the central region leads to smaller $R$ and vice versa. Furthermore, the value of reflectance $R$ depends not only on the optical parameters of the phantom but also on the collection angle $\alpha$ or NA of the imaging system. These can be quantitatively analyzed in Fig. 5 by plotting $R(x, y)$ along the
Figure 3: The two-layer skin model with a pigmented lesion at the center. Optical parameters of each layer or region: $(\mu_a 1, \mu_s 1, g 1, n 2)$ for epidermis; $(\mu_a 2, \mu_s 2, g 2, n 2)$ for dermis and $(\mu_a 3, \mu_s 3, g 3, n 3)$ for the lesion.

Figure 4: Three gray-scale reflectance images $R(x, y)$ of a semi-infinite heterogeneous phantom with a $201 \times 201$ grid over an FOV of 41.2 mm along $x$- and $y$-axis: (a) $\mu_a 3 = 2.00 \text{ mm}^{-1}$, $\alpha = 15.0^\circ$; (b) $\mu_a 3 = 2.00 \text{ mm}^{-1}$, $\alpha = 90.0^\circ$; (c) $\mu_a 3 = 0.15 \text{ mm}^{-1}$, $\alpha = 90.0^\circ$. Other parameters are $N_0 = 1.13 \times 10^8$, $r = 12.5 \text{ mm}$, $\theta_0 = 30^\circ$, $d = 0$, $D = 0.75 \text{ mm}$, $r = 4.00 \text{ mm}$. Other parameters are: $\mu_a 2 = 0.20 \text{ mm}^{-1}$, $\mu_s 3 = \mu_s 3 = 4.00 \text{ mm}^{-1}$, $g_2 = g_3 = 80$, $n_{r 2} = n_{r 3} = 1.50$.

Figure 5: (a) The $x$-dependence of $R(x, 0)$ with $\mu_a 3 = 2.00 \text{ mm}^{-1}$ at different values of numerical aperture ($=\sin \alpha$) as marked; (b) the contrast versus the albedo $a_3$ of the central region with $D = 0.75 \text{ mm}$ for different albedo of the peripheral region: $a_2 = 0.870$ and $\mu_t 2 = 6.32 \text{ mm}^{-1}$. The scattering coefficient of the central region are marked in the figures with arrows indicating the $a_2$ values and the solid lines are visual guides. Other parameters are identical to those in Fig. 4.

$x$-axis for different NA. Because of the symmetry, we averaged the photon density along the $y$-axis over 2 rows of pixels on each side of the $x$-axis to reduce the fluctuation.

From Fig. 5(a), one can see that the reflectance increases as the collection angle $\alpha$ increases.
to 90°, in which all photons exiting from the phantom surface and within the circular area of the lens contribute to the reflectance image. But the relative differences in $R$ between the two regions remains similar, indicating the portion of the image information that is independent of the imaging system parameters. Based on these results, we define an image contrast $C$ to characterize the relative change in reflectance as

$$C = \frac{< R_c > - < R_p >}{< R_c > + < R_p >},$$

where $< R_c >$ is the reflectance averaged over a circle concentric with the central region and $< R_p >$ is the reflectance averaged over a concentric ring in the peripheral region of illuminated area outside of the central region. For a central region of radius $r = 4$ mm, the radius of the averaging circle for $< R_c >$ is 3 mm and the inner and outer radii of the averaging ring for $< R_p >$ is 5 and 11 mm, respectively. With the above definition, we fist investigated the effect of collection angle $\alpha$ and the results shows clearly that $C$ has a very weak dependence on $\alpha$. In the following results, we adopted $\alpha = 90°$ to reduce the output variance in the following results of MC generated $R(x, y)$.

To find the relations between the image contrast $C$ and phantom parameters, we carried out a large number of MC simulations with different sets of optical parameters $(\mu_a, \mu_s, g, n_r)$ between the two regions. Analysis of these numerical image data demonstrated an interesting relation in that $C$ depends mainly on the single-scattering albedo $a_3 = \mu_s / (\mu_a + \mu_s)$ in the central region relative to the albedo $a_2$ in the peripheral region when $a_2 = a_3$ and $g_2 = g_2$. These results are presented in Fig. 5(b) with the thickness of central region $D = 0.75$ mm. The thickness of the first layer $d$ was set to 0.

The above MC model has been used to generate multispectral reflectance images for development of a principal-component based multivariate algorithm for analysis of the in vivo data [2]. These results will be presented.

REFERENCES