RGB Imaging System for Monitoring of Skin Vascular Malformation’s Laser Therapy

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ABSTRACT

A prototype RGB imaging system for mapping of skin chromophores consists of a commercial RGB CMOS sensor, RGB LEDs ring-light illuminator and orthogonally orientated polarizers for reducing specular reflectance. The system was used for monitoring of vascular malformations (hemangiomas and telangiectasias) therapy.

Keywords: RGB imaging, hemoglobin, skin vascular malformations, laser therapy monitoring, multi-spectral imaging

1. INTRODUCTION

Vascular malformation treatment is usually evaluated by visual inspection of dermatologist. Several studies have been performed for quantitative assessment of therapy by laser speckle imaging [1], optical coherence tomography [2], pulsed photothermal radiometry [3], diffuse reflectance spectroscopy and spectral imaging [4, 5]. Skin color image analysis has been offered as simple and fast method for treatment evaluation [6, 7].

Multi-spectral imaging can be used for non-invasive assessment of skin chromophore, e.g. oxy-/deoxy-hemoglobin and melanin, distribution; it reflects the overall skin condition and may facilitate better pathology diagnostics [8, 9]. However, commercial multi-spectral imaging cameras are bulky and expensive, so limiting their clinical implementation. Color digital RGB camera can be regarded as a low-cost alternative. It acquires three spectral (red (R), green (G) and blue (B)) images simultaneously, therefore can be regarded as a simple and fast spectral imaging device. In combination with specific narrow-band spectral light sources, RGB imaging could become competitive for some specific applications, including the assessment of hemoglobin concentration [10-12].

A prototype RGB imaging system for mapping and monitoring of hemoglobin distribution in skin has been designed and tested. The previous study showed that system can be used for monitoring of hemoglobin concentration changes during specific provocations – arterial/venous occlusions and heat test [12]. The aim of this study was to approbate clinically the RGB imaging system for evaluation of vascular malformations’ laser treatment results. The multi-spectral imaging system was used as a reference [8].

2. MATERIALS AND METHODS

2.1 Data acquisition

RGB imaging system consists of a commercial RGB CMOS sensor (USB uEye LE form IDS Imaging Development Systems), RGB LED ring-light illuminator and orthogonally orientated polarizers for reducing specular reflectance. The system was adapted with stand-off for working distance of 3 cm, providing a field of view 3x3 cm and spatial resolution 0.05x0.05 mm per pixel.

All settings were set to manual mode and maintained constant between measurements. The spacer of RGB device touched the skin around malformation during the measurements. The system was adapted for flat surface imaging. Measurements were taken in a dark environment to reduce artifacts induced by ambient lightning. One measurement lasted less than 10 seconds – most of the time was spend for positioning and less than a second for taking a snapshot.
The multi-spectral imaging system was used as the reference. It consisted of multi-spectral imaging camera (Nuance from Cambridge Research & Instrumentation), halogen lamp ring-light and also orthogonally orientated polarizers for reducing specular reflectance. The system was adjusted for spatial resolution 0.75x0.75 mm per pixel and spectral resolution 10 nm (bandwidth of the Nuance liquid crystal tunable filter). The spectral operating range was 450 – 950 nm [8]. The measurement lasted several minutes – most of the time was spend for positioning and ~ 1 minute for acquisition of spectral image cube.

2.2 Data processing

RGB camera acquires three spectral images at a time where R channel can be roughly attributed to 600…700 nm spectral range, G – 500…600 nm and B – 400…500 nm. RGB imaging system spectral sensitivity was calculated as the product of spectral sensitivity of camera and emission spectrum of the light source [12].

The acquired reflectance images (a set of intensity signals I) were transformed to optical density (OD) images that represent absorption difference compared to the reference:

$$\Delta OD = - \log \left( \frac{I}{I_{\text{ref}}} \right), \quad (1)$$

where $I_{\text{ref}}$ are the reference values taken from the surrounding normal skin area.

Optical density OD spectrum can be expressed as a superposition of chromophore’s spectra, so the OD difference in skin can be expressed as:

$$\Delta OD = \left( \varepsilon_{HbO_2} \cdot \Delta c_{HbO_2} + \varepsilon_{Hb} \cdot \Delta c_{Hb} + \varepsilon_{Mel} \cdot \Delta c_{Mel} \right) \cdot d, \quad (2)$$

where $\varepsilon_{OH}$, $\varepsilon_{DOH}$, $\varepsilon_{Mel}$ are molar extinction coefficients and $\Delta c_{OH}$, $\Delta c_{DOH}$, $\Delta c_{Mel}$ are concentration difference between point of interest and normal skin of the most significant skin chromophores – oxy-hemoglobin (HbO2), deoxy-hemoglobin(Hb) and melanin (Mel), d – optical path length, proportional to the light penetration depth in skin. Since difference values ($\Delta OD$ and $\Delta c$) are used, and melanin concentration changes between different points of image are not expected, influence of this chromophore can be neglected at Eq. (2).

$\Delta OD$ is obtained for each color (R, G and B), thus three equations can be used to calculate chromophore concentration changes and can be expressed in matrix form:

$$\begin{bmatrix}
\Delta OD(R) \\
\Delta OD(G) \\
\Delta OD(B)
\end{bmatrix} =
\begin{bmatrix}
\varepsilon_{HbO_2} \cdot (R) \cdot d(R) & \varepsilon_{Hb} \cdot (R) \cdot d(R) & 1 \\
\varepsilon_{HbO_2} \cdot (G) \cdot d(G) & \varepsilon_{Hb} \cdot (G) \cdot d(G) & 1 \\
\varepsilon_{HbO_2} \cdot (B) \cdot d(B) & \varepsilon_{Hb} \cdot (B) \cdot d(B) & 1
\end{bmatrix} \times
\begin{bmatrix}
\Delta c_{HbO_2} \\
\Delta c_{Hb} \\
\text{Offset}
\end{bmatrix}, \quad (3)$$

where $\Delta OD$ are values acquired from measurements, but concentration can be obtained fitting $\Delta c$ to experimental data. Offset value is included in equation system to compensate for intensity differences due to surface curvature.

Light penetration depth corrected absorption coefficients $\varepsilon \cdot d$ for R, G and B channels were calculated from tabular spectral data of molar extinction coefficients [13] and optical path length [14] as weighted values of RGB imaging system spectral sensitivity [12].

Total hemoglobin (tHb) concentration difference is expressed as a sum of oxy- and deoxy-hemoglobin concentration differences:

$$\Delta c_{tHb} = \Delta c_{HbO_2} + \Delta c_{Hb}. \quad (4)$$

As a result, three parameter distribution maps are obtained - HbO2, Hb and tHb. Data processing was performed by MatLab software. Processing time for one pixel was ~ 0.1 ms. VGA resolution (640x480 pixels) image processing time would be ~ 35 s.
A similar approach previously described by Jung et al [15] was used for quantitative analysis of skin lesions. The percent area ($S_{\text{threshold}}(\%)$) from parameter distribution maps was defined and computed for 9 different threshold levels:

$$S_{\text{threshold}}(\%) = \left( \frac{S_{\text{threshold}}}{S_{\text{total}}} \right) \cdot 100,$$

(5)

where $S_{\text{threshold}}$ represents the number of pixels above the selected threshold range and $S_{\text{total}}$ – the total number of pixels in the image.

To evaluate skin lesion responses to laser treatment, the percent change of $S_{\text{threshold}}(\%)$ ($\Delta S_{\text{threshold}}(\%)$) after therapy was computed for each threshold level:

$$\Delta S_{\text{threshold}}(\%) = \left( \frac{S_{\text{threshold}}(\%)_{\text{after}} - S_{\text{threshold}}(\%)_{\text{before}}} {S_{\text{threshold}}(\%)_{\text{before}}} \right) \cdot 100,$$

(6)

where $S_{\text{threshold}}(\%)_{\text{before}}$ and $S_{\text{threshold}}(\%)_{\text{after}}$ represent threshold area before and after laser therapy. Negative values correspond to decrease in the area of the skin lesion, but positive values – increase.

### 2.3 Participants

20 Caucasian adults (23 to 57 years old) with different skin vascular malformations were recruited for the study at The Clinic of Laser Plastics. 12 hemangiomas, 7 telangiectasias and 1 superficial vein network were dislocated at different sites of the body. All subjects studied had Fitzpatrick skin type II. 1064 nm Nd:YAG laser (160 J/cm² for 15…35 ms) was used for treatment of 17 cases and 810 nm diode laser (40 J/cm²) – for the rest 3 cases. Type and parameters of therapy device were selected by dermatologist after visual inspection of the lesion. The clinical trial was approved by the Local Ethics Committee.

Vascular malformations were inspected before, directly after and at least month after laser treatment by use of RGB and multi-spectral imaging systems.

### 3. RESULTS

Figure 1 shows a typical vascular malformation telangiectasia (#2) that appears darker and reddish compared to surrounding tissue. Distinctly darker spot (#1) can be observed in the center of malformation. Optical density spectra (Fig. 1a) indicate that increased absorption corresponds to hemoglobin characteristic bands between 500 and 600 nm.

Redness of the skin increases in lesion (#4) and surrounding tissue (#5) after the laser therapy (1064 nm Nd:YAG laser, 160 J/cm², 15 ms). Optical density spectra indicate increased hemoglobin concentration. The center of telangiectasia (#3) appears darker and changes color to gray or black. Blood vessels are destroyed and melted in this region. The shape of optical density spectrum changes significantly, increased absorption in spectral range between 600 and 800 nm was observed.

Acquired RGB images before, directly after and a month after laser treatment with the corresponding chromophore (tHb, HbO₂ and Hb) distribution maps are presented in Figure 1b. Telangiectasia appears as light spot in total hemoglobin maps before treatment and gradually disappears after the treatment. Total hemoglobin map can be attributed to erythema index and used for redness quantitative evaluation. Oxygenation drop in the destroyed central area of malformation and increase in the surrounding skin were observed after laser treatment. Thus the oxy-hemoglobin (HbO₂) map highlights active regions of the skin.

Total hemoglobin (tHb) concentration difference was chosen for quantitative analysis of skin lesions. Nine threshold values were selected in experimental evaluation of tHb amplitude. First threshold value was chosen 0, as zero indicates reference (normal skin). Each next value was increased by the step 0.0025. The percent area ($S_{\text{threshold}}(\%)$) and the percent change of $S_{\text{threshold}}(\%)$ ($\Delta S_{\text{threshold}}(\%)$) after therapy were calculated for each threshold value. Four lesions failed
quantitative analysis due to artifacts caused by specific dislocation (nose and close to lips). These subjects weren’t taken into further analysis despite visual changes in chromophore maps before and after treatment could be detected.

![Figure 1. Telangiectasia: (a) optical density difference spectra before and directly after laser treatment, (b) RGB images before, directly and a month after laser treatment and corresponding chromophore (tHb, HbO2 and Hb) distribution maps.](image)

The percent area for the first threshold level 0 ($S_1(\%)$) was in the range 30…80% which was much higher than the visually observed. As the skin is inhomogeneous and could have values above the reference value, this threshold level was not used for lesion quantitative analysis.

$S_2(\%)$ was in the range 1…5% of the total area and corresponded well with the visually observed sizes of vascular malformations. All other $S_{threshold}(\%)$ values were less than 1%. Therefore the second threshold $S_2(\%)$ was selected as the most appropriate for lesion quantitative analysis.

The percent change of $S_{threshold} (%)$ ($\Delta S_{threshold} (%)$) after therapy for 1-3 threshold levels is shown in Figure 2. $\Delta S_2(\%)$ has negative values for all 16 cases indicating decreased area of vascular malformation. The results corresponded well with our visual observation. As it was expected, $\Delta S_1(\%)$ was not stable and showed positive values in five cases. $\Delta S_3(\%)$ showed decrease in most cases.

Lesion No. 7 was superficial vascular malformation and did not change much after the laser therapy. Lesion No. 8 was hemangioma and improvements were minimal after the first treatment.

Comparing to the previous study done by Jung et al [15], the correlation between lesion areas before and after treatment ($R^2 = 0.40$) was relatively low. This might be explained taking into account that the present study covered different lesions and parameters of therapy.
4. CONCLUSIONS

RGB imaging system for mapping and monitoring hemoglobin changes in skin has been tested for evaluation of vascular malformations’ laser treatment. Skin hemoglobin maps obtained from RGB images corresponded well with those based on data of the multi-spectral imaging camera.

Total hemoglobin maps like the erythema index maps can be used for quantitative evaluation of skin lesion recovery. Oxy-hemoglobin map represents oxygenation and specifies the irritation zones.

The percent change of $S_{\text{threshold}}$ (%) after therapy proved to be a sensitive parameter for quantitative analysis of skin lesions. Since most of skin vascular malformations are dislocated on face, the method should be further improved for use on curved surfaces.

The advantages of RGB imaging are high performance and low equipment cost. This study shows that RGB imaging could become competitive for evaluation of vascular malformations’ laser treatment results. Simplicity of parameter computing allows nearly real-time measurements. From the other hand, advantage of conventional multi-spectral imaging is better spectral resolution.

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