# 3x3 Technique for RGB Snapshot Mapping of Skin Chromophores

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**Abstract:** Three monochromatic spectral images have been extracted from a single RGB image data set at simultaneous illumination of skin by 473nm, 532nm and 609nm spectral lines. They were further transformed into distribution maps of three skin chromophores - melanin, oxy-hemoglobin and deoxy-hemoglobin, related to pigmented and vascular skin malformations. Performance and clinical potential of the proposed 3x3 technique is discussed.
**OCIS codes:** (170.3880) Medical and biological imaging; (170.1870) Dermatology; (170.6935) Tissue characterization; (110.4234) Multispectral and hyperspectral imaging; (100.2000) Digital image processing; (170.0110) Imaging systems; (170.2945) Illumination design.

## 1. Introduction

Fast and reliable imaging of spectral reflectance is an important issue at a range of application areas including skin chromophore mapping for express-diagnostics, recovery monitoring, forensic evaluation, etc. [1-6]. The shorter is acquisition time of the spectral image data set, the lower is probability of image motion artefacts to be corrected afterwards. From the other hand, the narrower are the selected spectral bands, the higher is performance and reliability of spectral imaging and the subsequent parametric mapping. Therefore single snapshot monochromatic spectral imaging at several fixed wavelengths seems to be a promising technology for fast and reliable parametric imaging of skin with potentially wide clinical applications.

Snapshot narrowband spectral imaging by means of additional dispersive elements in front of the image sensors have been reported previously [7,8]. This approach, however, excludes direct use of conventional RGB image sensors that are widely available in consumer cameras, mobile phones, computers, web cameras, etc. RGB snapshot chromophore mapping based on skin color analysis without any additional elements was demonstrated in [9]. Drawback of this approach is complicated processing procedure that involves conversion of the tri-stimulus values of the CIE *XYZ* color system into the corresponding RGB values and solution of complicated integral equations comprising skin chromophore absorption bands. Such processing is time/resource-consuming and very sensitive to the measurement and calculation errors.

Alternatively, extraction of several spectral images from a single RGB image data set by exploiting the spectral features of the image sensor and/or spectrally specific illumination of the target have been proposed and discussed recently [10-14]. Snapshot RGB mapping of skin hemoglobin at dual-wavelength laser illumination was first demonstrated in [10]; performance of this technique can be further improved by the corresponding RGB crosstalk corrections [12]. As concluded in [14], uniform illumination simultaneously at three fixed wavelengths theoretically allows mapping the spectral reflectance of the target (e.g. skin) at each of these wavelengths by processing of a single RGB image data set. Solutions of relatively simple linear equations are suitable for image processing in this case, which is advantageous compared to [9] from the point of accuracy and rational use of time and computing resources.

This concept (provisionally called “3x3 technique”) was experimentally validated in the presented study for snapshot mapping of three main skin chromophores - melanin, oxy-hemoglobin and deoxy-hemoglobin. Monochromatic images at the illumination wavelengths 473nm, 532nm and 609nm were successfully extracted from single RGB image cubes of color targets and in-vivo skin with pigmented and vascular lesions. The obtained set of spectral images was further converted into the distribution maps of three above-mentioned skin chromophores.

## 2. The concept

Basic concept of the proposed technique was described in details earlier [14]. Generally, if the target illumination spectrum comprises only a limited number of narrow spectral lines with fixed wavelengths, monochromatic spectral images at these wavelengths can be extracted from a single snapshot RGB image data set by separate registration of the R, G and B output values from each image pixel. Subsequent mapping of the spectral reflectance at each of the illumination wavelengths is possible this way, if the following pre-conditions are met:

(i) the RGB spectral sensitivity curves of the image sensor are known - given by the manufacturer or directly measured,
(ii) they are uniform - the same for all pixels over the whole area of the RGB image sensor,
(iii) linear photovoltaic responses at all three detection channels (R, G and B) are ensured,
(iv) the image field comprises a white reference reflector, related to a particular pixel group of the image.

Mapping of spectral reflectance at three chosen wavelengths (473nm, 532nm, 659nm) was the first goal of this study. Spectral reflectance $k_\lambda$ is defined as the ratio $I(\lambda)/I_o(\lambda)$, where $I(\lambda)$ is the intensity of radiation reflected from the target at the specified wavelength $\lambda$, and $I_o(\lambda)$ - the intensity of radiation, reflected at this wavelength from a non-absorbing white reference, located at same distance from the photo-detector. If the pre-conditions (i)-(iv) are met, relatively simple linear equations describe RGB digital output signals of the image sensor [14]. The main variables influencing the results are ratios of the spectral sensitivities of each detection band (R, G and B) at the three defined wavelengths and the measured output signals of the R-, G- and B-channel from the pixels located in the target zone and in the white reference zone. In result, crosstalk-corrected spectral reflectance images at the three specified wavelengths were constructed.

The second goal was to convert the obtained set of three monochromatic images into distribution maps of skin melanin, oxy-hemoglobin and deoxy-hemoglobin. Three-chromophore skin model [4,5] based on the Beer-Lambert law leads to the following system of 3 equations to be applied for each x-y pixel of the three spectral images:

$$\begin{cases} c_a \cdot \varepsilon_a(\lambda_1) + c_b \cdot \varepsilon_b(\lambda_1) + c_c \cdot \varepsilon_c(\lambda_1) = -\frac{\ln k_1}{l_1} \\ c_a \cdot \varepsilon_a(\lambda_2) + c_b \cdot \varepsilon_b(\lambda_2) + c_c \cdot \varepsilon_c(\lambda_2) = -\frac{\ln k_2}{l_2} \\ c_a \cdot \varepsilon_a(\lambda_3) + c_b \cdot \varepsilon_b(\lambda_3) + c_c \cdot \varepsilon_c(\lambda_3) = -\frac{\ln k_3}{l_3} \end{cases} \quad (1),$$

where $k_i$ – spectral reflectance at the wavelength $\lambda_i$ ($\lambda_1 = 473$nm, $\lambda_2 = 532$nm, $\lambda_3 = 659$nm), $l_i$ – wavelength-dependent mean optical path length, $\varepsilon(\lambda_i)$ – extinction coefficient of the specified chromophore (a – oxy-hemoglobin, b – deoxy-hemoglobin, c - melanin) at the i-wavelength, $c_j$ – relative concentration of the chromophore to be mapped (j = a, b or c).

Tabulated values of spectrally-dependent extinction coefficients of melanin and hemoglobin [15,16] at the three specified wavelengths were used in calculations. As for the absorption path lengths, several options have been tested with respect to the skin anatomy and different mean light penetration depths $\delta$ in the skin [17]. The 473nm radiation ($\delta \sim 190\mu$) can be absorbed mainly by skin melanin within the epidermal layer of thickness 150-200 $\mu$ [18], but also some absorption by the hemoglobin of upper dermal blood vessels has to be taken into account. The two other exploited wavelengths (532nm, $\delta \sim 328\mu$ and 609nm, $\delta \sim 655\mu$) are both notably absorbed by epidermal melanin and dermal hemoglobin. Our analysis showed that reasonable results can be obtained with estimated path length $4\delta$ in all cases but for 473nm absorption by hemoglobin, where $l = 2\delta$ appeared to be good approximation.

Solution of the system (1) for each image pixel at the above-mentioned conditions and embedded numerical data concerning the nine extinction coefficients [15,16] is:

$$c_a = 0.145 \cdot \ln k_3 - 0.033 \cdot \ln k_2 - 0.024 \cdot \ln k_1 \quad (2),$$
$$c_b = -0.128 \cdot \ln k_3 - 0.024 \cdot \ln k_2 + 0.057 \cdot \ln k_1 \quad (3),$$
$$c_c = -0.738 \cdot \ln k_3 + 0.408 \cdot \ln k_2 - 0.806 \cdot \ln k_1 \quad (4).$$

Expressions (2-4 ) were further used for mapping skin oxy-hemoglobin, deoxy-hemoglobin and melanin, respectively.

### 3. Measurement setup

To validate the above-proposed model, series of RGB image measurements from a color target and human skin were taken. The imaging setup comprised an objective-supplied CMOS camera (*USB 2 UI-1226LE-C*, IDS with known RGB sensitivity curves [19]) and three stabilized cw laser modules (473 nm, MBL-III-473; 532 nm, DD532-10- 5; 659 nm, DB650-12-3(5)) with comparable output powers in the range 10…15 mW. Each laser output beam was launched in a SMA-terminated bundle of 7 fibers with silica core diameter 400 microns; their outputs formed an illumination ring (composed of 21 randomly distributed emitting fiber ends) around the camera objective. Every laser module could be independently switched on and off. The target plane was located 50mm from the CMOS objective lens. Uniformity of target illumination was ensured by a toroidal lens in front of the emitter ring. Polarizing film was placed behind the lens, and a film with orthogonal polarization direction – in front of the camera objective. Laser output powers and the exposition time of CMOS sensor were carefully adjusted to provide linearity of photovoltaic response at all spectral combinations used in the experiments. All automatic features of the CMOS

image sensor were switched off. Flat paper target composed of four color segments (red, green, blue and white) was used for primary calibration of the system, like in the previous study [14]. The colors were selected to assure a unique spectral image pattern at each illumination wavelength that allowed comparing the "true" spectral images (under illumination of a single laser line) with those extracted from the RGB image data set under simultaneous illumination by three different laser lines. The illumination and recording conditions were optimized with respect to such comparison. After that, the color target was replaced by healthy or pathological in-vivo skin areas in order to map the distributions of three skin chromophores.

## 4. Results

Fig.1illustrates the obtained chromophore distribution maps for a pigmented skin nevus. One can see increased melanin content in the malformation area and practically unchanged oxy- and deoxy-hemoglobin content. In the cases of vascular pathologies notable changes in oxy- and deoxy-hemoglobin content were recorded instead.

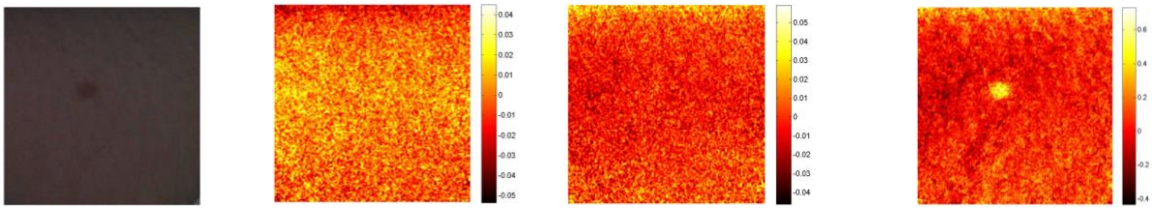

Fig.1. Left to right: skin nevus image, oxy-hemoglobin map, deoxy-hemoglobin map and melanin map.

The obtained results demonstrate snapshot mapping of three mayor skin hromophores by conventional RGB image sensors using the proposed 3x3 technique. They confirm the technical challenge to apply consumer digital RGB cameras for fast and reliable express-diagnostics of skin by adding an adjusted poly-chromatic illumination source and appropriate software. Direct use of all registered R, G and/or B pixel output signal values with subsequent corrections of the spectral RGB inter-channel crosstalk is a fast, cost-efficient and robust way to compose spectral images in situations when spectrally specific target illumination is available. The proposed approach may be further extended for parametric mapping of other diagnostic criteria of skin and other bio-tissues.

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