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RGB IMAGING DEVICE FOR MAPPING AND MONITORING OF HEMOGLOBIN DISTRIBUTION IN SKIN

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A prototype RGB imaging device for the mapping and monitoring of hemoglobin distribution in skin was designed and tested. The device was examined for monitoring hemoglobin concentration changes during specific provocations: arterial/venous occlusions and heat test. Besides, hemoglobin distribution maps of rosacea on a cheek were obtained.

Keywords: RGB imaging, hemoglobin, skin

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1. Introduction

The distribution of skin chromophore, e. g. oxy-/deoxy-hemoglobin and melanin, can be non-invasively assessed by use of multi-spectral imaging [1, 2]. However, commercial multi-spectral imaging cameras are bulky and expensive, thus limiting their clinical implementation.

A colour digital camera can be regarded as an alternative. It acquires three spectral (red (R), green (G) and blue (B)) images simultaneously, therefore can be regarded as a simple and fast multi-spectral imaging device. In combination with specific narrow-band spectral light sources, RGB imaging could become competitive for some specific applications, including assessment of hemoglobin concentration [3–5].

A prototype RGB imaging device for the mapping and monitoring of hemoglobin distribution in skin was designed and tested in this work. The principle that was used in our previous study [2] for the assessment of chromophore concentration by use of multi-spectral imaging was adapted for RGB imaging.

2. Experiment

The designed device consists of a commercial RGB CMOS sensor (USB uEye LE Imaging from *IDS Development Systems GmbH* [6]), RGB LED ring-light illuminator, and orthogonally orientated polarisers for reducing specular reflectance [7]. The system was adapted for the working distance of 3–10 cm. The camera was operated in two modes: "photo" for a single image acquisition, and "video" for dynamic process observation.

The RGB camera works as a simple multi-spectral imaging device acquiring three spectral images at a time where R channel can be roughly attributed to the 600–700 nm spectral range, G to 500––600 nm, and B to 400–500 nm (Fig. 1(a)). The RGB imaging device spectral sensitivity (Fig. 1(b)) was calculated as a product of spectral sensitivity of the camera and spectral intensity of light source (Fig. 1(a)).

The acquired reflectance images I_R, I_G, I_B were transformed into optical density *OD* images that represent absorption changes compared to the reference:

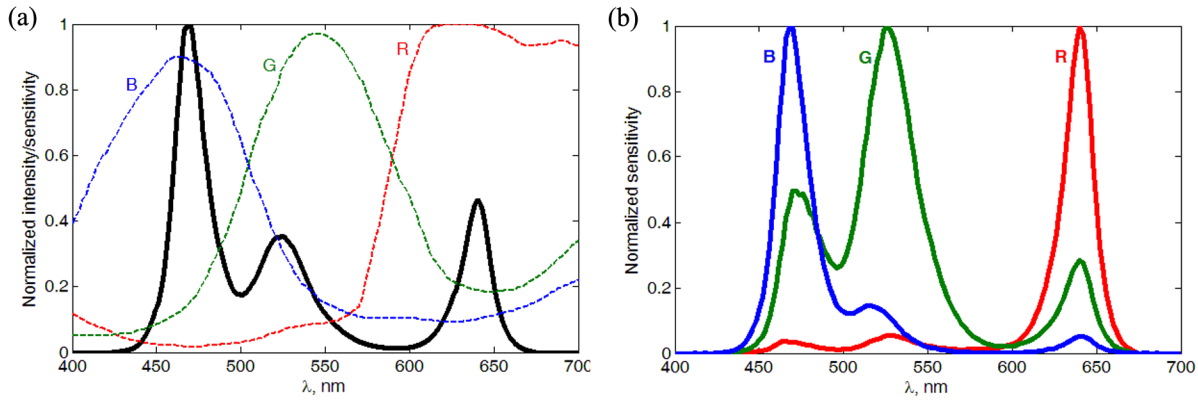

Fig. 1. (a) Normalised RGB LEDs spectrum (full line) and R, G, and B channel spectral sensitivity (dotted line). (b) Normalised RGB imaging device spectral sensitivity.

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

where I_{ref} are the reference values taken from a normal skin area before provocation.

Optical density *OD* spectrum is a superposition of chromophore's spectra, so *OD* changes in skin can be expressed as

$$\Delta OD = (\varepsilon_{\text{OH}} \cdot \Delta c_{\text{OH}} + \varepsilon_{\text{DOH}} \cdot \Delta c_{\text{OH}} + \varepsilon_{\text{Mel}} \cdot \Delta c_{\text{Mel}}) \cdot d, \quad (2)$$

where ε_{OH}, ε_{DOH}, ε_{Mel} are molar extinction coefficients and Δc_{OH}, Δc_{DOH}, Δc_{Mel} are concentration changes of typical skin chromophores: oxy-hemoglobin (OH), deoxy-hemoglobin (DOH) and melanin (Mel), d is optical path length proportional to the light penetration depth in skin. If specific chromophore concentration changes are not expected (e. g. melanin in the present study), this chromophore can be ignored in Eq. (2).

Δ*OD* is obtained from measurements for each colour, thus three equations can be used to calculate chromophore concentration changes. The above-determined system of simultaneous linear equations can be expressed in a matrix form:

$$\begin{bmatrix} \Delta OD(\text{R}) \\ \Delta OD(\text{G}) \\ \Delta OD(\text{B}) \end{bmatrix} = \begin{bmatrix} \varepsilon_{\text{OH}}(\text{R}) \cdot d(\text{R}) & \varepsilon_{\text{DOH}}(\text{R}) \cdot d(\text{R}) \\ \varepsilon_{\text{OH}}(\text{G}) \cdot d(\text{G}) & \varepsilon_{\text{DOH}}(\text{G}) \cdot d(\text{G}) \\ \varepsilon_{\text{OH}}(\text{B}) \cdot d(\text{B}) & \varepsilon_{\text{DOH}}(\text{B}) \cdot d(\text{B}) \end{bmatrix} \times \begin{bmatrix} \Delta c_{\text{OH}} \\ \Delta c_{\text{DOH}} \end{bmatrix}. \quad (3)$$

The light penetration depth corrected absorption coefficients $\varepsilon\, d$ for R, G and B channels were calculated from tabular spectral data of molar extinction coefficients [8] and optical path length [9] as weighted values of the RGB imaging system spectral sensitivity (Fig. 1(b)). For the current RGB imaging device, Eq. (3) can be expressed as

$$\begin{bmatrix} \Delta OD(\text{R}) \\ \Delta OD(\text{G}) \\ \Delta OD(\text{B}) \end{bmatrix} = \begin{bmatrix} 8.98 & 21.49 \\ 46.83 & 49.48 \\ 32.49 & 26.78 \end{bmatrix} \times \begin{bmatrix} \Delta c_{\text{OH}} \\ \Delta c_{\text{DOH}} \end{bmatrix}, \quad (4)$$

where Δ*OD* are values acquired from measurements, but concentration can be obtained by fitting Δ*c* to experimental data. The total hemoglobin (TH) concentration change is a sum of OH and DOH concentration changes:

$$\Delta c_{\text{TH}} = \Delta c_{\text{ON}} + \Delta c_{\text{DON}}. \quad (5)$$

For monitoring hemoglobin dynamic changes during provocation, the region of interest (RoI) is determined and the parameter is calculated as a mean value over the whole region.

The device was examined for monitoring hemoglobin concentration changes during specific provocations: arterial/venous occlusions and heat test. Besides, hemoglobin distribution maps of rosacea on a cheek were obtained.

3. Results and discussion

Occlusion tests are often used to study system response to changes of OH and DOH in time [3].

In this work, a cuff was applied to the upper arm (170–180 mmHg for arterial and 90–100 mmHg for venous occlusion), but measurements were taken from the dorsal side of the hand: 20 s for reference acquisition of unprovoked skin, 1 min during occlusion, and 70 s for post-occlusive reaction observation. Parameters were calculated as mean values over the RoI.

As expected, OH concentration decreased, but DOH concentration increased due to oxygen starvation during arterial occlusion (Fig. 2(a)). The opposite reaction was observed after cuff release (OH rapid increase and DOH decrease) resulting also in TH overshoot. The TH increase during occlusion appeared due to a slow pressure increase in the cuff. The results corresponded well with the results obtained in other studies [3].

During venous occlusion (Fig. 2(b)) blood congestion as DOH and TH increased during provocation was expected and also observed.

The heat test was performed to cause local vasodilatation of blood vessels. A middle finger was immersed in warm water (40–50 °C) for 0.5 min and reaction was observed for 2 min (Fig. 3). Two RoIs were taken for calculation of OH, DOH and TH concentration changes: one from a provoked area and the other from a non-provoked finger. The TH and OH concentration increase was observed in the provoked area, but no significant

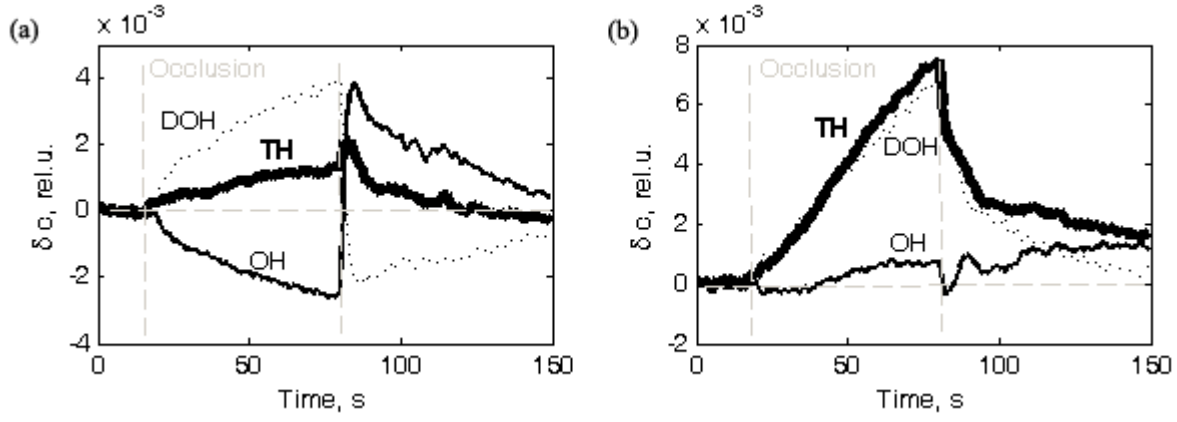

Fig. 2. Changes of oxy-haemoglobin (OH), deoxy-haemoglobin (DOH) and total haemoglobin (TH) concentrations during (a) arterial and (b) venous occlusion.

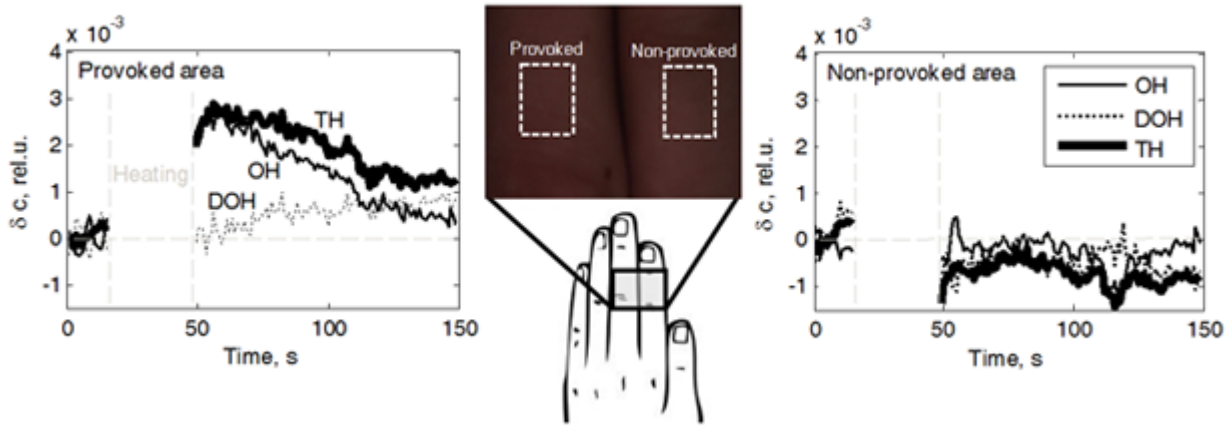

Fig. 3. (left) Changes of oxy-haemoglobin (OH), deoxy-haemoglobin (DOH) and total haemoglobin (TH) concentrations during heat provocation. (right) Reference signal from non-provoked area.

changes were found in the non-provoked side. The heat test measurement showed that the "video" mode can be used for monitoring parameter dynamic changes at several RoIs.

The colour images of rosacea on a cheek and its OH, DOH, and TH concentration maps are shown in Fig. 4. Rosacea is a vascular malformation, so an increased hemoglobin level compared to the normal surrounding tissue was expected and observed there. The best contrast between rosacea and the rest of the skin appeared at OH maps with increased OH level. The opposite nature was observed at DOH maps – DOH concentration decreased. The "photo" mode is suitable for mapping skin chromophores providing spatial information.

4. Conclusions

The RGB imaging device for the mapping and monitoring of hemoglobin changes in skin was designed and tested. The device can distinguish between both types of hemoglobin (oxy- and deoxy-hemoglobin) and can be used for the observation of dynamic processes like occlusions and vasodilatation, as well as for the chromophore concentration mapping of vascular malformations, thus confirming its potential for skin diagnostics. The simplicity of parameter computing allows real-time measurement implementation.

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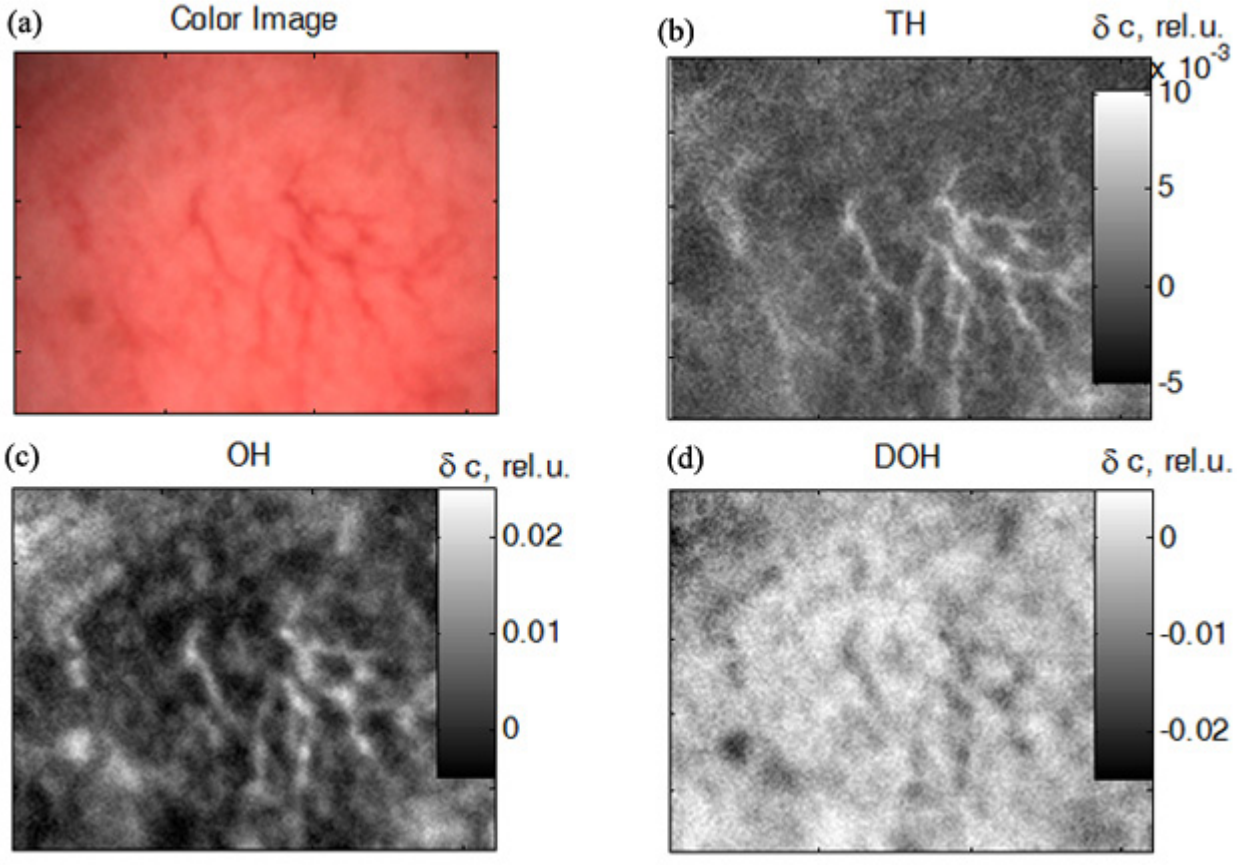

Fig. 4. (a) Colour image of rosacea on a cheek, and relative concentration maps of (b) total haemoglobin, (c) oxy-haemoglobin and (d) deoxy-haemoglobin distribution.

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TRISPALVIO (RGB) VAIZDINIMO ĮRENGINYS, SKIRTAS NUSTATYTI IR STEBĖTI HEMOGLOBINO PASISKIRSTYMĄ ODOJE

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Santrauka

Sukurtas ir išbandytas prototipinis trispalvio (*angl.* red-green-blue, RGB) vaizdinimo įrenginys, skirtas nustatyti ir stebėti hemoglobino pasiskirstymą odoje. Ištirta, kaip prietaisu galima stebėti hemoglobino koncentracijos pokyčius, kai yra konkrečios juos sukeliančios priežastys – užsikimšę arterijos ar venos bei kraujagysles praplečiantis šiluminis testas. Be to, gauti hemoglobino pasiskirstymo atvaizdai skruosto rozacėjos atveju.