Optical skin assessment based on spectral reflectance estimation and Monte Carlo simulation

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ABSTRACT

Optical non-contact measurements in general, and chromophore concentration estimation in particular, have been identified to be useful tools for skin assessment. Spectral estimation using a low cost hand held device has not been studied adequately as a basis for skin assessment. Spectral measurements on the one hand, which require bulky, expensive and complex devices and direct channel approaches on the other hand, which operate with simple optical devices have been considered and applied for skin assessment. In this study, we analyse the capabilities of spectral estimation for skin assessment in form of chromophore concentration estimation using a prototypical low cost optical non-contact device. A spectral estimation workflow is implemented and combined with pre-simulated Monte Carlo spectra to use estimated spectra based on conventional image sensors for chromophore concentrations estimation and obtain health metrics. To evaluate the proposed approach, we performed a series of occlusion experiments and examined the capabilities of the proposed process. Additionally, the method has been applied to more general skin assessment tasks. The proposed process provides a more general representation in form of a spectral image cube which can be used for more advanced analysis and the comparisons show good agreement with expectations and conventional skin assessment methods. Utilising spectral estimation in conjunction with Monte Carlo simulation could lead to low cost, easy to use, hand held and multifunctional optical skin assessment with the possibility to improve skin assessment and the diagnosis of diseases.

Keywords: spectral estimation, skin assessment, chromophore concentration, occlusion measurment, optical, non- contact

1. INTRODUCTION

Skin assessment is usually performed by visual examination by a physician. The diagnosis depends on the subjective judgement of the physician and the skin samples have to be extracted for further investigation of the health status. Optical measurements, on the other hand, could provide objective non-invasive examination. Hence these techniques could avoid scarring and pain for the patient during the diagnoses. Skin colorants like melanin, oxygenated hemoglobin and deoxygenated hemoglobin called chromophores and their concentrations can provide useful information about the health status of skin. This research addresses chromophore concentration estimation and mapping with a prior proposed skin assessment device based on spectral estimation in combination with Monte Carlo simulation.

The *SkImager* proposed by Spigulis et al.[1] is a low cost non-contact optical measurement device for skin assessment. It has been proposed designed and tested prior to this study, but will be a tool of investigation for this research. Jakovels et al.[2] used a similar technology for monitoring of vascular lesion phototherapy efficiency.

In this research we shall develop and investigate the capabilities of the *SkImager* for spectral estimation of skin reflectances. The estimates shall be used for skin assessment in general and skin chromophore estimation in particular, furthermore the process shall be extended with priori Monte Carlo simulated diffuse reflectance spectra.

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Processor	Nvidia Tegra 2 T20 a dual-core ARM Cortex-A9 processor (clock frequency 1 GHz)
Sensor	RGB CMOS, 3Mpix (2048 x 1536 pixels), Pixelsize: $3.2\mu m^2$ (MT9T031)
Storage	Removable SD card
Display	4.3 inch 480 x 272 pixel touchscreen
Dimensions	121 x 205 x 101 mm
Weight	˜440g

Table 1. Technical Data of the *SkImager* and the Aptina CMOS sensor

2. THEORETICAL BACKGROUND AND THE *SKIMAGER*

The *SkImager* is a previously proposed prototypical compact device for skin assessment. It was developed in the Biophotonics Laboratories in Riga Latvia and is described in detail in a previous publication.[1] A round skin spot illuminated with 5 polarized narrow band LED's can be imaged by a cross oriented polarized CMOS Sensor. The illuminations covers the VIS (visible) and IR (infra red) spectrum with narrow band LED's at $450nm$, $540nm$ $660nm$ and $940nm$ and a white LED. The controls of the *SkImager* are realized in form of a touchscreen on the back of the device. All parts are assembled in a 3D printed housing.

2.1 Spectral Sensitivity

The spectral sensitivity of the sensor is an important factor for the spectral estimation the data sheet sensitivities were taken to evaluate the coverage of spectral sensitivity along the visual spectral band. The spectral sensitivity

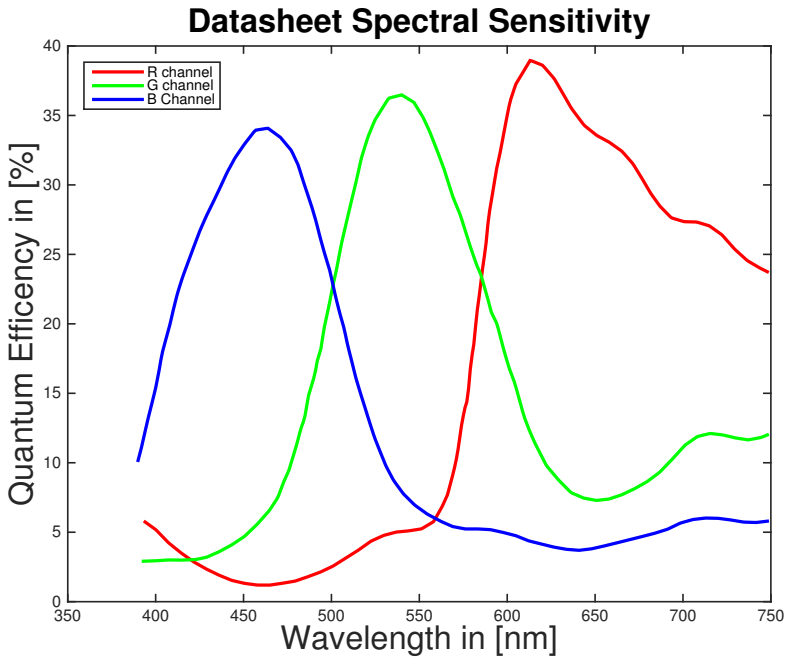

Figure 1. Spectral Sensitivity of the sensor in *SkImager* according to data sheet.

given by the manufacturer were used and are presented in Figure 1. These curves are an important factor of a sensor of an imaging system used for spectral estimation.

Both color quality and spectral estimation results are connected to the spectral sensitivity. Another very important spectral feature of an imaging system are the spectral power distributions of the LEDs discussed in the following section.

2.2 Spectral Power Distribution of the LEDs

The Spectral Power Distribution of the Illumination is an important aspect for modeling the reflectance of a sample with known spectral reflectance. To measure the spectral power distribution of the *SkImager* its

illumination was directed towards (0°) a reference white and the reflected light was measured with an *Avantes AvaSpec-ULS2048*. The measurement fibre was directed in a 45° angle towards the reference white to avoid specular reflections of the light source. Following the CIE norm (0°/45°) discussed in[3]

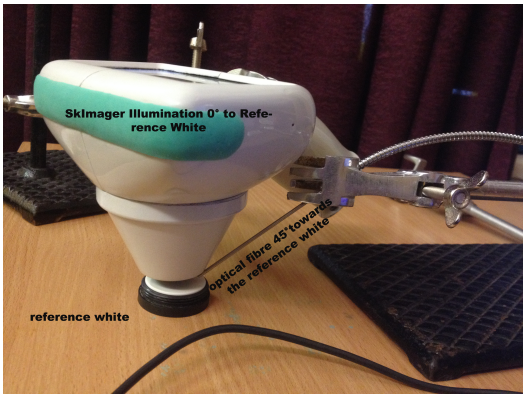

Figure 2. Measurement setup to measure the spectral power distribution of the LEDs in the *SkImager* with a 0°/45° measurement setup according to the CIE[3]

The *Avantes AvaSpec-ULS2048* with a spectral range of 350nm to 1100nm and a spectral resolution of 0.5nm was used. To account for the dark current and ambience light during the measurement a black measurement was taken and subtracted from the measurements. All measurements were repeated three times and averaged to decrease effects of random noise. Figure 3 shows the measured spectral power distributions of the illumination

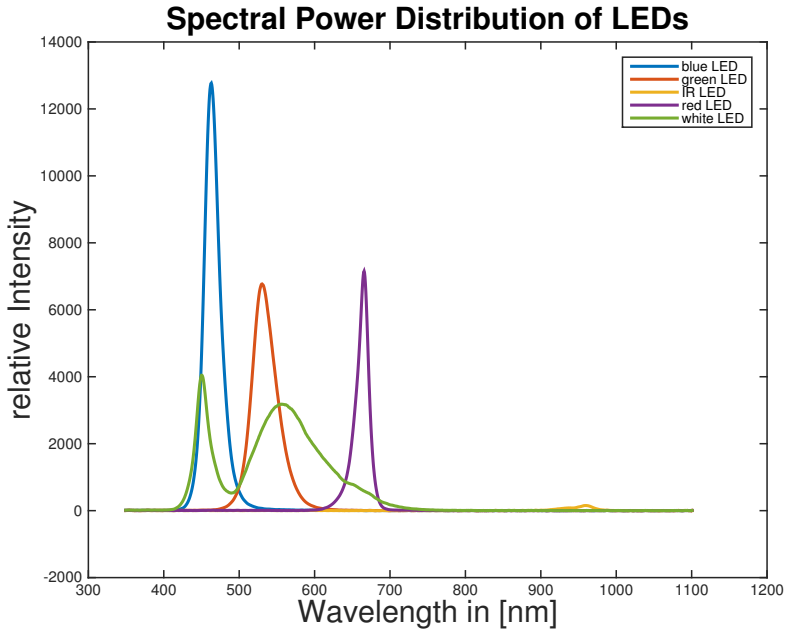

Figure 3. Relative measurement of the LED spectral power distribution measured as described in Section 2.2 and

in a relative measurement scale of the device. Figure 4 on the other hand shows a normalized spectrum ranging from $0 to 1$.

To obtain the normalized spectra each value of all curves was divided by the curves maximum value. We can clearly see that the relative power of the IR LED was considerably lower than the other LEDs. The Blue LED provides by far the highest output.

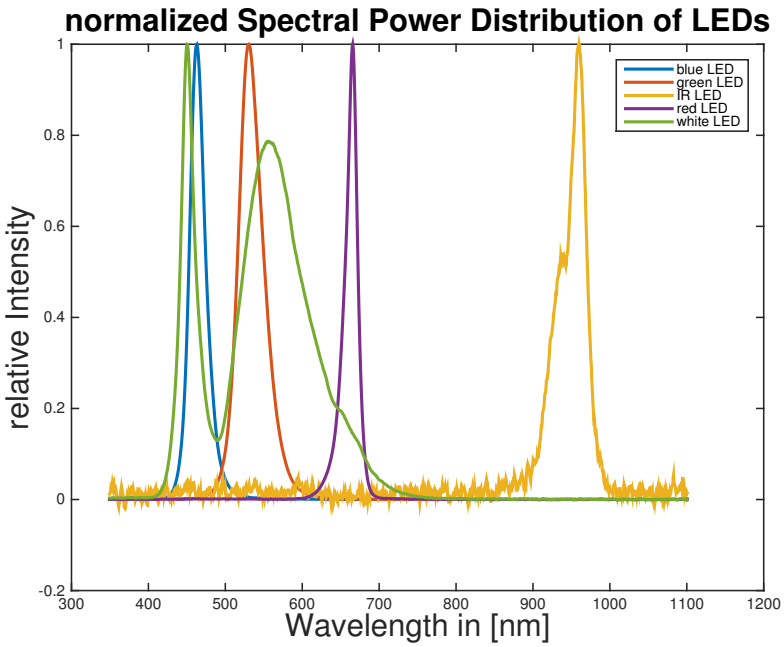

Figure 4. normalized LED spectral power distributions normalized by dividing with the highest value of each curve

2.3 Effective Spectral Sensitivity

In order to compute the effective spectral sensitivity per channel of the *SkImager* we multiplied the spectral power distribution of the LEDs and the spectral sensitivity of the sensor. The Figure 5 shows the effective sensitivity

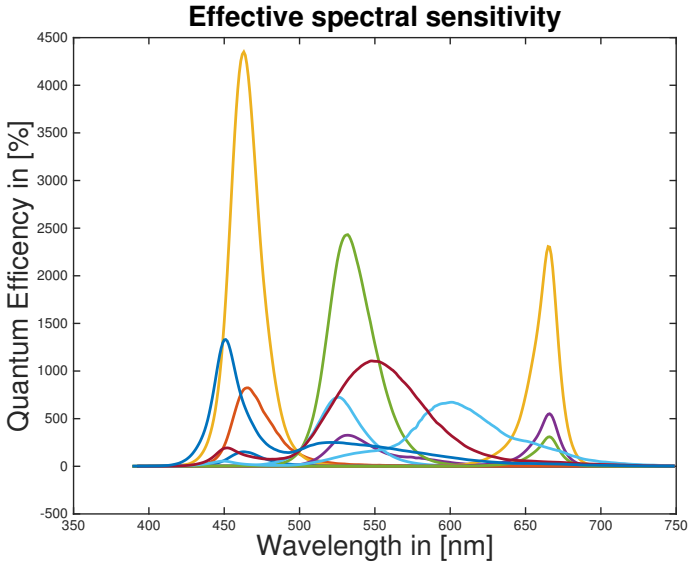

Figure 5. Effective spectral sensitivity computed by multiplication of SPD of the LEDs and spectral sensitivity of the Sensor

per channel in the visual range of the spectrum from (400 to 700nm) and shows that it is not evenly distributed over the spectral range. Infrared region of the spectrum was left out in this study also due to the fact that the manufactures didn't provide the spectral sensitivity of the sensor in this region. For spectral measurements or spectral estimation it is desirable to have an uniform spectral sensitivity over the whole range of the spectrum.

Both benefit from a spectrally uniform illumination with enough signal along the whole spectral band of interest. Especially,the signal is weak around 500nm were none of the illuminations provides an adequate output of energy.

3. METHODS

3.1 Monte Carlo Simulation

Prior to the spectral estimation we performed a series of Monte Carlo simulations using MCML by Wang et al.[4] For the configuration of the MCML simulation we followed the published code by Atencio et al.,[5] defining a 3 layer model. With an epidermis, dermis and subcutaneous tissue layer. In total we simulated four different sets of diffuse skin spectra changing the concentration of the dominant chromophore. One set of simulations with dominant oxyhemoglobin and different concentrations, one with different overall concentrations of blood with a constant oxygenation level, one set with dominant but varying bilirubin concentration and one set with dominant but varying melanin concentration.

In the case of oxyhemoglobin we changed the oxygen saturation of the blood from 0.20 - 0.70 in 0.10 steps, while keeping blood volume fraction constant (0.02 mg/L), melanin concentration constant(0.02 mg/dl) and a low level of bilirubin (0.1g/L) constant through the simulations. For the simulation with different volume

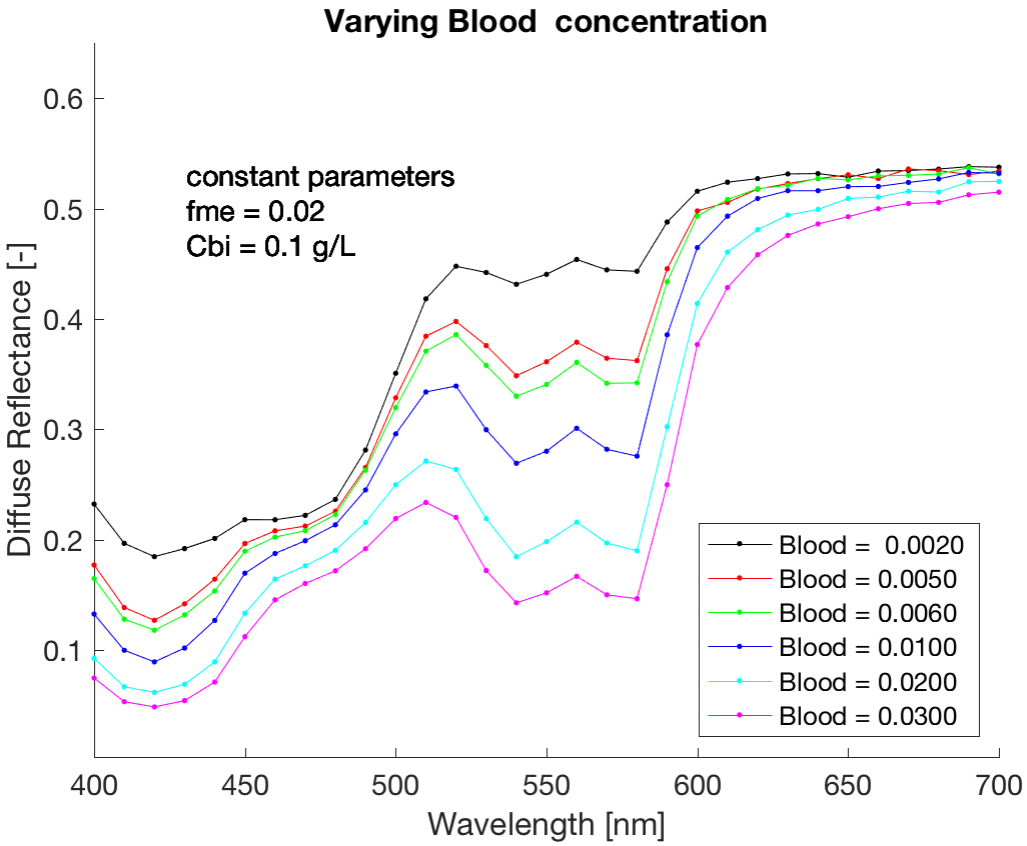

Figure 6. example of simulated spectra in this case with varying total blood concentration and constant bilirubin, oxygen saturation, and melanin based on plots MCML simulation code and theory proposed by Atencio et al.[5]

fractions of blood we kept the bilirubin concentration (0.1 g/L), melanin concentration (0.02) and the oxygen saturation (0.70) as constant, while varying the blood volume fraction to 0.002, 0.005, 0.006, 0.01, 0.02, 0.03. The plot for different blood concentrations can be seen in Figure 6.

Furthermore one set with changing the total bilirubin concentration (0 0.05 0.1 0.15 0.20 0.25) while keeping melanin (0.02), blood volume (0.02) fraction and oxygen saturation (0.70) constant was simulated. And similarly one set with varying melanin concentration keeping blood, oxygenation and bilirubin constant.

3.2 Linear Least Square Fitting in Lower Dimensional Space

The spectral estimation we choose is a linear least square fitting in a lower dimensional space also known as the *Imai Berns* Method.[6] Lower dimensional reflectance factors are obtained by performing dimensionality reduction on a set of training reflectances. The method requires a prior taken training set with corresponding training reflectances. Principal component Analysis (PCA) is then performed on the training data set to obtain the linear base and the training set coefficients. This method is according to the authors[6] more robust to noisy channels as a result of the optimization in a lower dimensional space. We then obtain a full spectral image cube with an estimated spectrum for each pixel in the image.

3.3 Chromophore estimation

The estimated spectra were used to estimate chromophore concentrations per pixel which can then be visualised using chromophore heat maps. An overview of the flow chart of the proposed method can be seen in

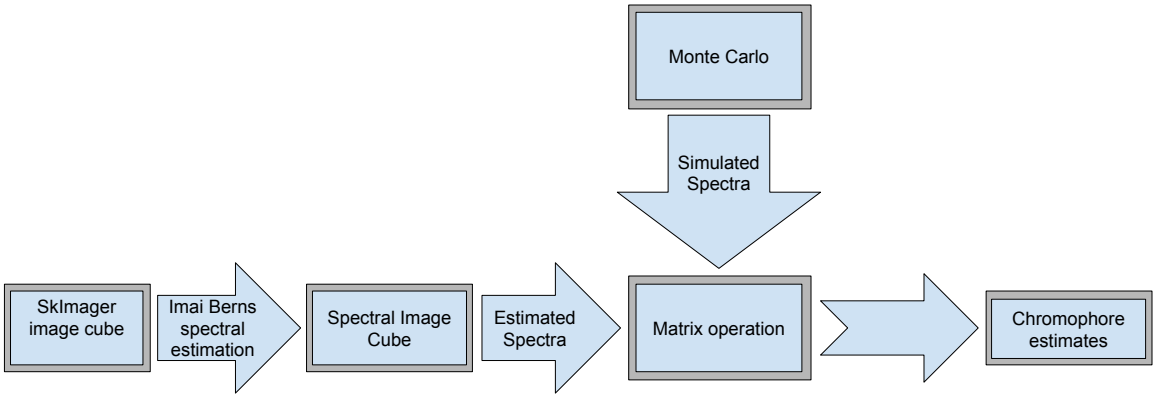

Figure 7. Flowchart of the proposed method, spectral estimations based on Imai Berns[6] method are used in conjunction with Monte Carlo simulations to obtain chromophore estimates

Figure 7. The estimated spectra are used in conjunction with the simulated spectra to estimate chromophore concentrations. For the further investigation usually an average of a pixel mask was used to account for noise. A main consideration for the estimation of the chromophore concentration was computationally efficiency and robustness. Following the limitations of the *SkImager* hardware and to ensure a work flow with instantaneous results. All the development of algorithms was performed in *Matlab*. Considering computational efficiency in *Matlab* matrix operations are well suited. A simple matrix model was formulated to estimate the chromophore concentrations based on the Monte Carlo simulations with a dominant chromophore,

$$\tilde{CC}_{sxn} = C_{pxs} * \tilde{A}^{\top}_{nxs} \tag{1}$$

where CC is the concentration of each chromophore per pixel, C is a matrix with the different Monte Carlo simulated spectra with p different dominant chromophores and $\tilde{A}_{hxs}$, $\tilde{A}_{hxs}$ is the estimated absorption spectrum for each pixel (h) resulting from the *SkImager*.

3.4 Occlusion Measurement

In order to verify the performance of the proposed method. We performed the occlusion test often used to study diffusion of tissue.[7–10] The occlusion test requires consecutive measurements of the hand of a patient who's arm is clamped with an inflatable cuff. Inflating the cuff blocks the incoming flow of blood and simultaneously stops the flow of blood out of the hand. The occlusion test is a well known study with a known outcome and therefore a suitable proof of concept measurement to verify the performance of the proposed algorithm. The concentration

of oxygenated Hemoglobin in the hand should fall exponentially during the occlusion (starting from the point of cuff inflation).[7–10] Deoxygenated hemoglobin increases exponentially during the occlusion.[8–10]

In total 12 volunteers (4 female and 8 male) all caucasian skin type with 2 of the male subjects with darker skin and an age distribution from age $22-34$ years were measured in a time frame of 5 minutes. The measurement setup was chosen to minimize the effects of specular reflection for the spectrophotometer measurement and to minimize the time between the two measurements. A white LED was used to take the spectral measurements as a result of the limited space for the measurement setup. The measurement geometry was $0°/45°$ with the

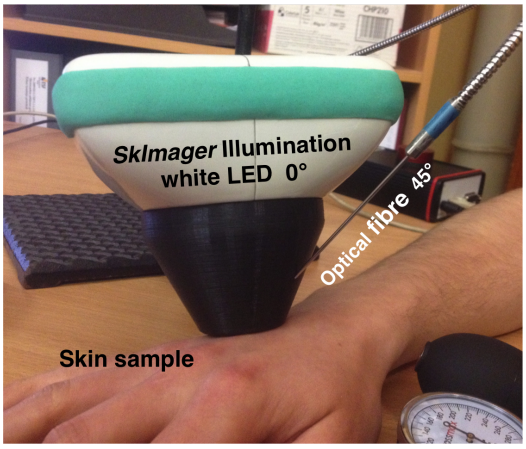

Figure 8. Measurement setup to measure the skin training set with ground truth reflectance measurement using *Avantes Avaspec* and the corresponding *SkImager* responses of the LEDs *SkImager* white LED illumination with a $0°/45°$ measurement Setup according to the CIE[3]

light source normal to the sample and the detector in a 45° angle to avoid specular reflections according to the CIE[3] the setup can be seen in Figure 8. Using a white LED for the spectral measurements is not optimal but it was unavoidable for the limited space for the measurement setup. We can also consider the white LED as sufficiently uniform in the spectral band of interest ($450-650$nm). Considering the temporal arrangement of the measurement a single measurement with both devices took about 15 seconds where the SkImager imaging took $6-10$ seconds and the spectrophotometer measurement about 5 seconds.

4. RESULTS AND DICSUSSION

In the following paragraph we discuss results of the proposed method applied to occlusion measurements. The chromophore estimation described in Section 3.3 based on prior estimated spectra calculated with the colorchecker training using 6 channels and the *Imai Berns* method as discussed in Section 3.2 and Monte Carlo simulated skin spectra. Figure 9 shows the results for all subjects. All subjects chromophore concentration images were averaged and then combined into one average subject. Average estimate concentrations were then plotted over time. The curve shows the exponential decay of oxygenated hemoglobin saturation during the 2 minute period of occlusion. We can clearly see a good agreement of the general expected shapes or physiological behaviour which has been discussed in the literature.[7–10] The measurements were normed to have zero concentration for the first measurement by subtracting the average concentration of the first five measurements. Therefore, the plot shows relative changes over time compared to the base line measured prior to occlusion. Also the expected oxygen overshoot can be seen in the plots obtained through spectral estimation combined with monte carlo simulation.

5. CONCLUSION

In this research we analyzed an existing optical non-contact skin assessment device called *SkImager* and proposed a spectral estimation workflow for chromophore estimation.

A spectral estimation workflow has been implemented for the *SkImager*. The estimated spectral image cube were used to estimate chromophore concentrations. The main objective was hereby a computationally efficient implementation usable for the *SkImager* and with the possibility to operate in real time.

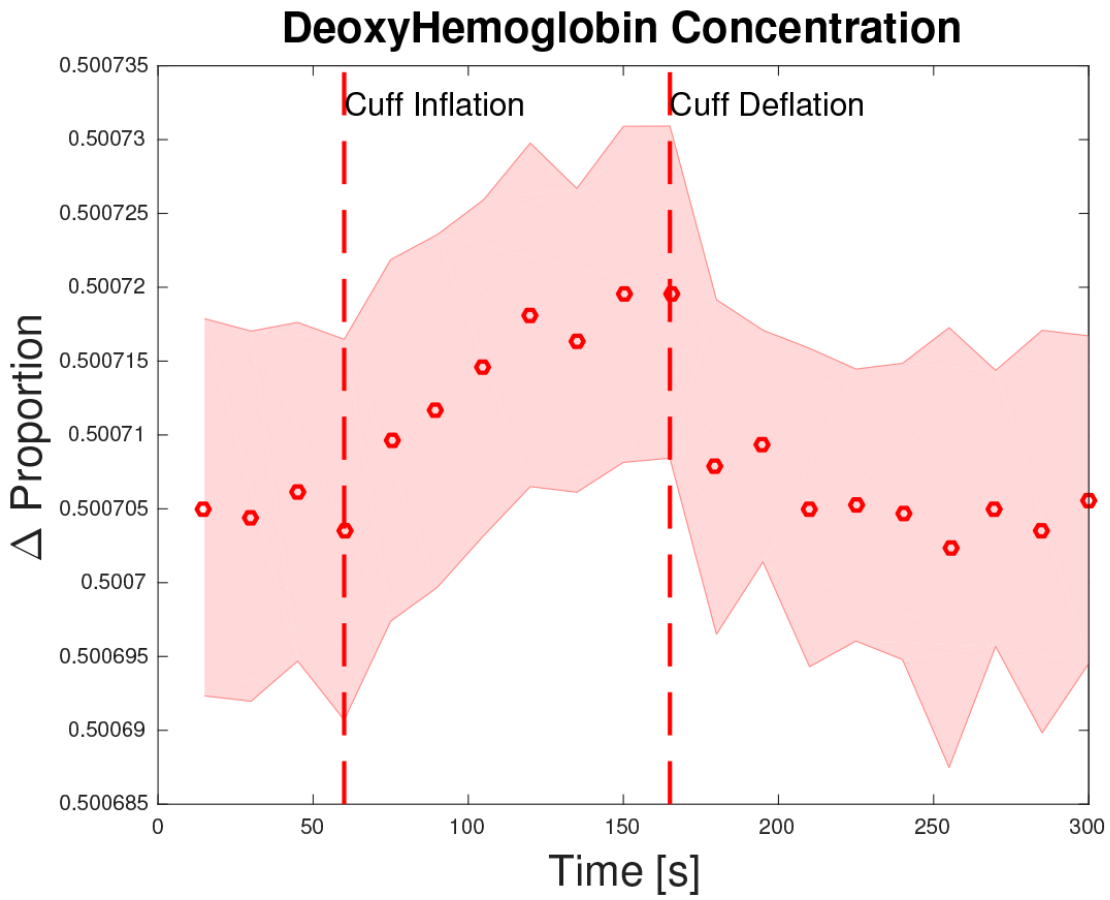

Figure 9. Average deoxygenated hemoglobin concentration of all patients averaged. Expected exponential increase during occlusion

Occlusion experiments were used to verify, chromophore concentration estimation in a realistic experiment. The results indicate that the proposed spectral estimation workflow combined with Monte Carlo simulation provides promising results and leads to expected oxygenation results for the occlusion measurement. Additionally the more general spectral image cubes provides full spectra and could be utilized for more complex analysis of the skin in the future.

Acknowledgment

Part of this work was supported by national funding from the University of Eastern Finland and by the Latvian National Research Program SOPHIS under grant agreement Nr.10-4/VPP-4/11 and the Rhones Alpes scholarship Xplora. Some of the work was conducted in the Biophotonics Laboratory division of the University of Latvia in Riga. I hereby acknowledge everybody at the Biophotonics Laboratory involved in this work and especially Prof. Janis Spigulis and the prior work on the *SkImager*.

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