Journal of Medical Imaging

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Abstract. We propose an architecture with a remote phosphor-based modular and compact light-emitting diode (LED) light source in a noncontact dermoscope prototype for skin cancer screening. The spectrum and color temperature of the output light can easily and significantly be changed depending on spectral absorption characteristics of the tissues being imaged. The new system has several advantages compared to state-of-the-art phosphor converted ultrabright white LEDs, used in a wide range of medical imaging devices, which have a fixed spectrum and color temperature at a given operating point. In particular, the system can more easily be adapted to the requirements originating from different tissues in the human body, which have wavelength-dependent absorption and reflectivity. This leads to improved contrast for different kinds of imaged tissue components. The concept of such a lighting architecture can be vastly utilized in many other medical imaging devices including endoscopic systems. [DOI: 10.1117/1.JMI.2.4.044501]

Keywords: biomedical imaging; remote-phosphor system; skin-cancer screening; contrast enhancement; digital dermoscopy.
Paper 15026RR received Feb. 10, 2015; accepted for publication Sep. 9, 2015; published online Oct. 8, 2015.

1 Introduction

Early detection is a key to successful treatment of melanoma skin cancer and other skin diseases. Digital dermoscopy is a standard method for the diagnosis of various skin diseases. It is a noninvasive visual investigation process, based on optical criteria to evaluate pathological changes, so that advanced digital dermoscopic techniques help in detecting skin diseases at an early stage and minimize unnecessary excision and biopsy of skin lesions or nevi. Prominent examples of visual diagnosis criteria of melanoma skin cancer are the so-called ABCD rule (asymmetry, border, color, diameter) and the seven-point checklist (see, e.g., Ref. 1). In this context, differences between the classical contact dermatoscopy and the advanced contactless dermoscopy are noteworthy. A short overview of both methods is given in the following two subsections.

1.1 *Contact Digital Dermoscopy*

The typical commercial contact-type dermoscope (CTD) is a reflected light microscope with additional lighting and is placed directly on the skin. It requires the application of index matching gels and human supervision for each and every nevus under test. Digital contact dermoscopes also have an integrated digital camera to display and archive the images acquired. In addition, modern state-of-the-art systems provide diagnostic software to assist the physicians. Some of the commercially available and clinically established digital contact dermoscope systems are the Dermoscope by FotoFinder, the MoleMax by Derma Medical Systems Handels und Entwicklungs GmbH, and the DermoGenius pro by Dermoscan GmbH. The Dermoscope and DermoGenius, aided by immersion fluid, operate with polarized light to suppress disturbing reflections of the skin surface. All three systems use ultrabright white LEDs with constant color temperature for lighting.

1.2 *Noncontact Digital Dermoscope*

Compared to a contact dermoscope, a noncontact digital dermoscope (NCDD)[1] can pave the way for fully automatized and touch-free screening (i.e., without using gels) of melanoma for the whole human body. This can expedite the screening procedure, especially for patients with a large number of nevi. Good image clarity, automated screening with NCDD, and software integration of the ABCD rule of melanoma detection can improve the speed and quality of melanoma screening in an economically viable manner. The standard ABCD criteria and seven-point checklist are applied to score a suspicious lesion. It has been observed that in case of a nonplanar or protruding nevus, the contact plate of the CTD may strongly affect the three-dimensional topology of the nevus and could result in a distorted image of the same. Such distortions can be totally avoided with NCDD; hence, more accurate imaging is feasible.[1] Note that pushing the CTD probe on certain skin lesions can be painful for the patient. Naturally, with a remote or noncontact dermoscope, the screening procedure is generally more agreeable for the patient and completely painless.

1.3 *Dermoscopy Systems with Changeable Illumination Spectra*

In complex nevi, different kinds of pigments and underlying structures have different optical properties and with the aid of diagnosis software in combination with illumination at different wavelengths, also details that go beyond the visual subjective dermoscopy, such as scattering properties of pathological tissue can be collected and extend the diagnostic abilities.

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Contemporary commercial dermoscopes like the Dermoscope, MoleMax, or DermoGenius mostly use high-power tungsten halogen or multiple white LED-based light sources. Moreover, there are systems such as the SIAscope[2] and MelaFind,[3] which illuminate the lesions with different wavelengths, matched to their specific absorption and scattering properties. White light images can be produced with the Melafind system by simultaneous illumination with as many as 10 different wavelengths. Kapsokalyvas et al.[4] of the University of Florence also developed a contact dermoscopy system with illumination that is specifically adapted to the diagnostic characteristics. In this system, an LED ring of red, green, and blue LEDs is used for individual or simultaneous illumination.[4] The simultaneous illumination of the three LEDs is used to produce white light images. Analyzing the spectra of the light sources can help in the postprocessing of the images obtained from typical CCD/CMOS cameras, especially if the postprocessing algorithms rely on color-based feature extraction. This process would facilitate imaging of different lesions with higher contrast. The drawback in the production of white light images from just a few wavelengths, as realized in the latter systems, is that in this case, the white light spectra have large spectral gaps compared to bright white LEDs. This significantly affects the imaging of constituents that have their maximum absorption within the gaps of the illumination spectrum. An interesting report on hyperspectral imaging with wavelength-selective light source is presented in Ref. 5.

Our original NCDD prototype, realized in a previous work, incorporates an ultrabright white LED source, projection optics, imaging optics including crossed polarizers, and a CCD camera.[1] LEDs are efficient, compact, and cost-effective and hence are preferred over many other light sources. LEDs also offer excellent lifetimes in the range of 20,000 to 50,000 h, which are unmatchable by conventional lamps.[6] However, the spectral limitations of a typical white LED, especially the "cool white" type, often affect the color rendering and image quality. As image quality is of utmost importance in digital dermoscopy as well as for many other biomedical imaging devices, the design of the corresponding light sources can also play a vital role in the overall performance.

Careful consideration of all these facts led us to the development of an advanced NCDD prototype in this work that utilizes a remote phosphor light source architecture, where the output spectra can be varied easily in a very controlled and cost effective manner.

2 Novel Light Source for Application in a Noncontact Digital Dermoscope

The typical phosphor converted (PC) white LEDs are made of a single or multiple blue (~460 nm) LED chip(s) with phosphor (e.g., Ce:YAG) coating on top. A part of the blue photons emitted from the chip gets absorbed by the phosphor and downconverted in energy to generate a wideband spectrum ranging from green to red. Hence, the residual or unabsorbed blue photons and this downconverted wide spectrum together generate the overall "white" light output. This is schematically shown in Fig. 1.

In this work, the same principle of white light generation used in the PC LEDs is utilized in the remote phosphor architecture too, the only difference being a physically separable phosphor module instead of a fixed and "proximate" phosphor coating. The remote phosphor architecture, a promising approach in the field of general lighting applications, offers many benefits.[6] The physical separation of the blue LED chip and the phosphor module facilitates thermal management. The phosphor module is not exposed to high temperature as on the surface of the blue LED chip; therefore, it could offer better long-term reliability in some cases. Note that remote phosphor modules are reported to have a lifetime in the order of 50,000 h.[7] On the other hand, remote phosphor design can enhance the output luminous efficacy by up to 60% as compared to standard PC LEDs.[6] This, of course, involves careful design of the mixing chamber incorporating the blue LED chip and the remote phosphor module. The design of the mixing chamber also plays a significant role in determining the correlated color temperature (CCT) of the output light, for a given spectrum of the blue LED source and the remote phosphor concerned. The nominal CCT values mentioned for commercial remote phosphor modules are valid only for a given blue LED wavelength (e.g., ~460 nm) and a reference mixing chamber design. Hence, it should be noted that any variation in the blue source wavelength or mixing chamber design will result in deviations from the nominal conversion efficacy and CCT. However, a detailed treatment of the general technicalities of remote phosphor architecture is beyond the scope of this article.

Discrete RGB LEDs can also be used for good contrast enhancements for skin imaging purposes.[4] However, light field homogenization and full spectral coverage are typically challenging in such cases, unless the skin target is very close to the light sources (e.g., handheld dermoscope) or additional optics for RGB beam homogenization are utilized. Typically, each of the RGB LED chips offers a spectral full width half maximum of about 20 to 30 nm. Hence, they cannot cover the entire region of the white light spectrum obtained from PC LEDs. On the other hand, the use of xenon lamps would consume more space and generate more heat. One has to be careful with possible ultraviolet (UV) radiation from xenon lamps as well. Also, the CCT cannot be changed without changing the

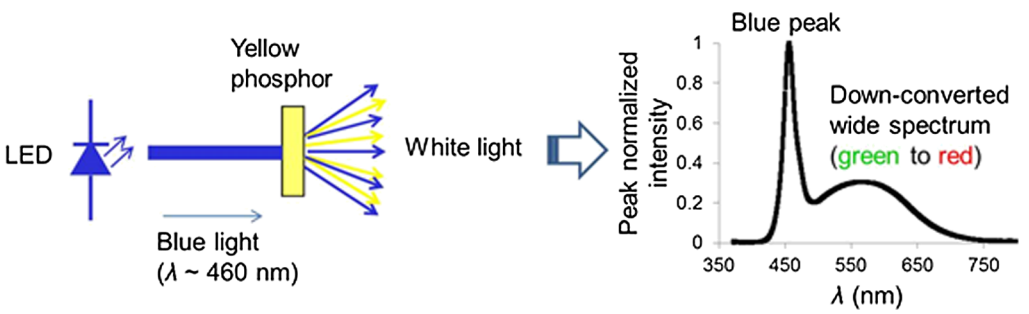

Fig. 1 Schematic of white light generation from a remote phosphor module irradiated by a blue LED.

xenon lamp itself. In addition, driver requirements are much more complicated for xenon lamps than in case of LEDs.

Note that the remote phosphor architecture is not limited to the use of blue LEDs as primary sources and blue laser diodes (~450 nm) can also be used for high luminance applications.[8] One very interesting application for a laser-irradiated phosphor source could be in surgical endoscopic systems.[9] With careful design of such a system, the CCT of the output white light can be varied in order to see particular tissues with better clarity.

Figure 2 shows the new NCDD prototype with remote phosphor architecture. Unlike in our previous prototype,[1] which was equipped with an ultrabright white LED (Luminus CBT-90 White), a blue LED (Luminus CBT-90 Blue) has been used here along with a commercial remote phosphor plate. A mixing chamber is built around the blue LED and the remote phosphor in order to enhance and homogenize the output flux. This new design offers an enormous possibility in terms of spectral tuning by changing the remote phosphor plate itself. One can foresee a situation where a clinician or technician can easily replace one designated phosphor plate with another in order to change the spectra of the light source for different target requirements. Such phosphor plates can be as cheap as <1$, whereas the ultrabright white or blue LED can cost around 50 to 60$ for single pieces from third-party vendors/distributors. Hence, for example, replacement of a cool white LED with a warm white one may not be economically viable at all. Commercial remote phosphor plates with a variety of CCT profiles are available for general lighting applications. As a proof of concept, commercial phosphors designed for general lighting applications were sufficient enough for our first tests. However, in future applications with more precise and complex spectral design requirements, phosphor chemists can optimize the composition and tweak the resulting spectra accordingly to a great extent. One must carefully note that the strength of the system architecture proposed here lies in its simplicity and low-cost components. A much more sophisticated light source with a programmable multiwavelength feature could technically offer better detection efficacy, but such complex systems remain beyond the reach for many clinics simply due to significantly higher costs of ownership.

In Fig. 3, the output spectra obtained from two different remote phosphor modules (Ph1 and Ph2) in our prototype setup are compared. The operating current of the blue LED was kept fixed at 5.5 A in both cases. It is worth mentioning

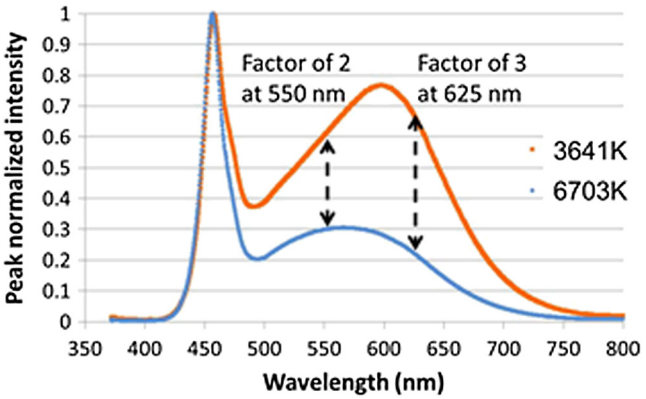

Fig. 3 The spectral comparison of the output white light from two different remote phosphor modules, as measured within an integrating sphere connected to a spectrometer. The correlated color temperature (CCT) values corresponding to the spectra are shown in the legend. The difference in output spectrum for different wavelengths is also indicated.

that variation in the current or temperature of the chip can change the output flux and CCT of the system. Such characteristics of the Luminus CBT-90 LED used in our prototype can be found in the product datasheet.[10] Therefore, the relevant operating parameters of the LED were kept constant for all measurements to ensure comparability. It is clearly seen that the transmitted spectrum of the blue LED that is used for phosphor excitation in both cases shows a sharp peak around 460 nm. However, the downconverted output spectra of the phosphors are significantly different. The measured CCT of Ph1 is 6703 K, whereas a CCT of 3641 K is measured with Ph2. In the output white light spectra, normalization to the pump peak shows that the Ph2 output spectrum is twice that of Ph1 at 550 nm and even three times stronger at 625 nm. This can have significant influence on the different contrast enhancement mechanisms described in Sec. 2. Note that the light spot luminance values at the target surface were measured to be 5780 lux and 4820 lux for Ph1 and Ph2, respectively.

The Commission internationale de l'éclairage (CIE) 1931 color space chromaticity diagrams in Fig. 4 show the coordinates of the white light output obtained from Ph1 [Fig. 4(a)] and Ph2 [Fig. 4(b)]. Clearly, Ph1 offers a bluish tint (cool

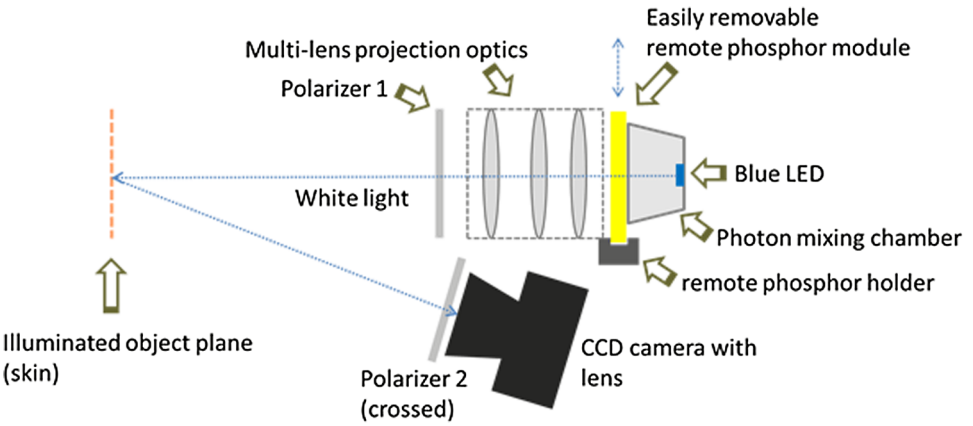

Fig. 2 Schematics of our noncontact digital dermoscope (NCDD) with replaceable remote phosphor-based white light source.

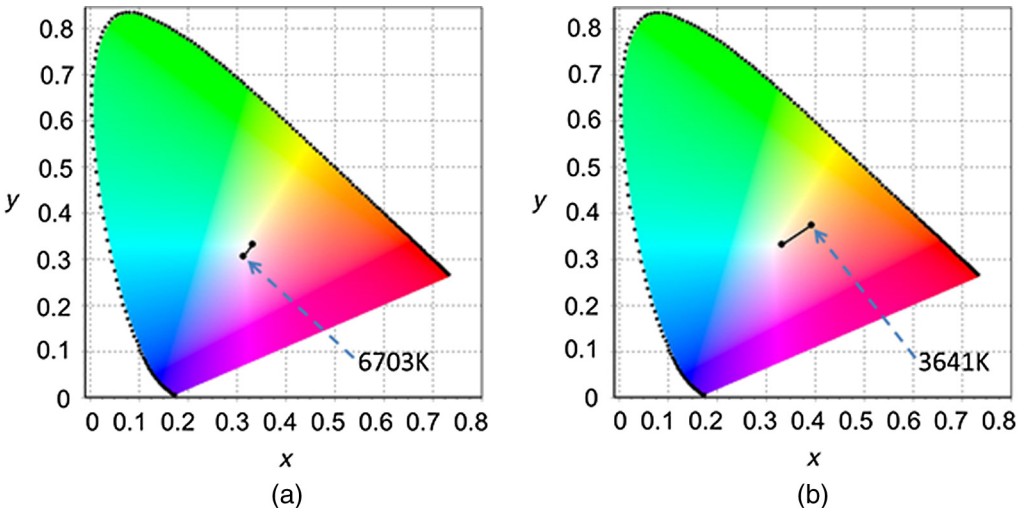

Fig. 4 The Commission internationale de l'éclairage (CIE) color charts showing the color coordinates corresponding to the two different remote phosphors used (a) Ph1 and (b) Ph2. The measured CCT values are denoted by the dashed arrows. The short black line to the measured point is drawn from the CIE 1931 reference white point.

white), whereas Ph2 renders a warmer white light, as expected from the spectra in Fig. 3.

3 Image Contrast Enhancement Tools and Influence of the Light Source Spectra

Postprocessing tools can enhance the image contrast of certain components of the nevi images. As far as human skin is concerned, there are several pigments with wavelength-dependent absorption characteristics. Hemoglobin and melanin are two of the best-known pigments in human skin. Depending on the nature and degree of a skin disease, the amount and spatial distribution of such pigments can differ significantly from normal skin conditions for the same individual. Hence, the underlying structures of a nevus can be significantly enhanced by using different contrast mechanisms[4] in the postprocessing of dermoscopic images. This can uncover features helpful for (early) diagnosis of certain skin diseases or monitor the changes in a nevus over time, which is particularly relevant for high-risk patients. In all these cases, the doctors would not only see the "normal" camera images, but also can possibly investigate some postprocessed enhancements for more structural details, which might otherwise go unnoticed. In this article, our discussion will be limited to blood contrast enhancement (BCE) and melanin contrast enhancement (MCE) mechanisms only.

3.1 *Blood Contrast Enhancement*

Blood contrast enhancement (BCE)[4] is a simple image processing technique used to enhance the visibility of vascular structures (blood vessels) in the papillary dermis of a nevus under investigation. This is favored by the fact that hemoglobin absorbs green light (e.g., 530 nm) much more than red light (e.g., 632 nm). Hence, a normalized subtraction of the green channel values from the red channel values can reveal vascularity in great detail as a part of the absorption by melanin in the normal skin background is suppressed in this way. The formula used for BCE on an image taken by our NCDD prototype with a remote phosphor source (Ph2) is schematically shown in Fig. 5. The resulting picture in false colors clearly indicates the enhancement of blood contrast. The variable multiplier shown in Fig. 5 is basically a gain-scaling factor for an 8-bit (0-255) image channel.

3.2 *Melanin Contrast Enhancement*

Melanin is the most dominant chromophore in human skin and is normally found in the upper layer of human skin, the epidermis.[4,11] The pigment is produced by melanocytes, a cell type situated at the junction between the epidermis and the underlying dermis and is contained within membranous particles known as melanosomes. In some cases, such as pigmented nevi, melanocytes and melanin are found in the dermis as well. Melanocytic nevi generally have higher melanosome contents than normal skin, which can be used to improve the contrast between the respective skin areas. Besides, data from animal models suggest that benign and malignant melanocytic nevi differ in their depolarization power, which could be due to increased melanin absorption of the tumor[11] or tumor morphology. The red channel has minimum blood contrast and hence is chosen to enhance the melanin contrast by a histogram rescaling mechanism, as shown in Fig. 6. Melanin has an almost exponentially decaying absorption spectrum from UV to near-infrared.[12] The absorption of both oxygenated hemoglobin (oxyHb) and deoxygenated hemoglobin (deoxyHb), however, is not as even but shows a significant minimum in the range from 600 to 750 nm.[13] Hence, having a strong intensity level in this range, e.g., at 625 nm, could be helpful while looking for melanin contrast regions in a nevus while minimizing the background absorption from oxyHb and deoxyHb.[13] Note that the choice of a feature-based "diagnostic window" shown in Fig. 6 can be suitably modified depending on the relative contrast between the melanin features and the surroundings (background).

Keeping in mind the wavelength-specific (or say R-G-B channel-specific) relative absorption levels in the aforementioned contrast mechanisms, as well as the Bayer filters used in typical color CCD/CMOS sensors, it is obvious that by tuning the spectral intensity distribution of the light source, one can enhance the contrast and clarity significantly. This could also mean that under certain conditions, the signal-to-noise ratio

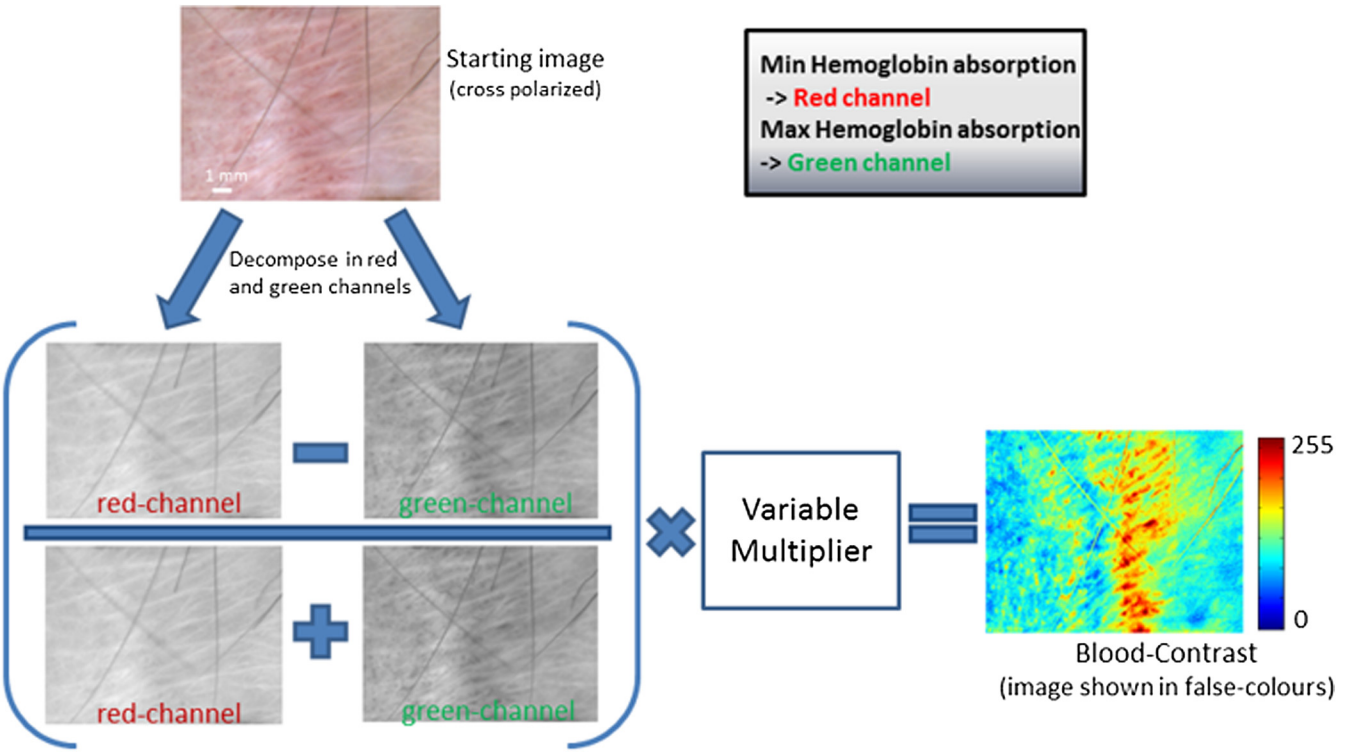

Fig. 5 The blood contrast enhancement (BCE) mechanism.

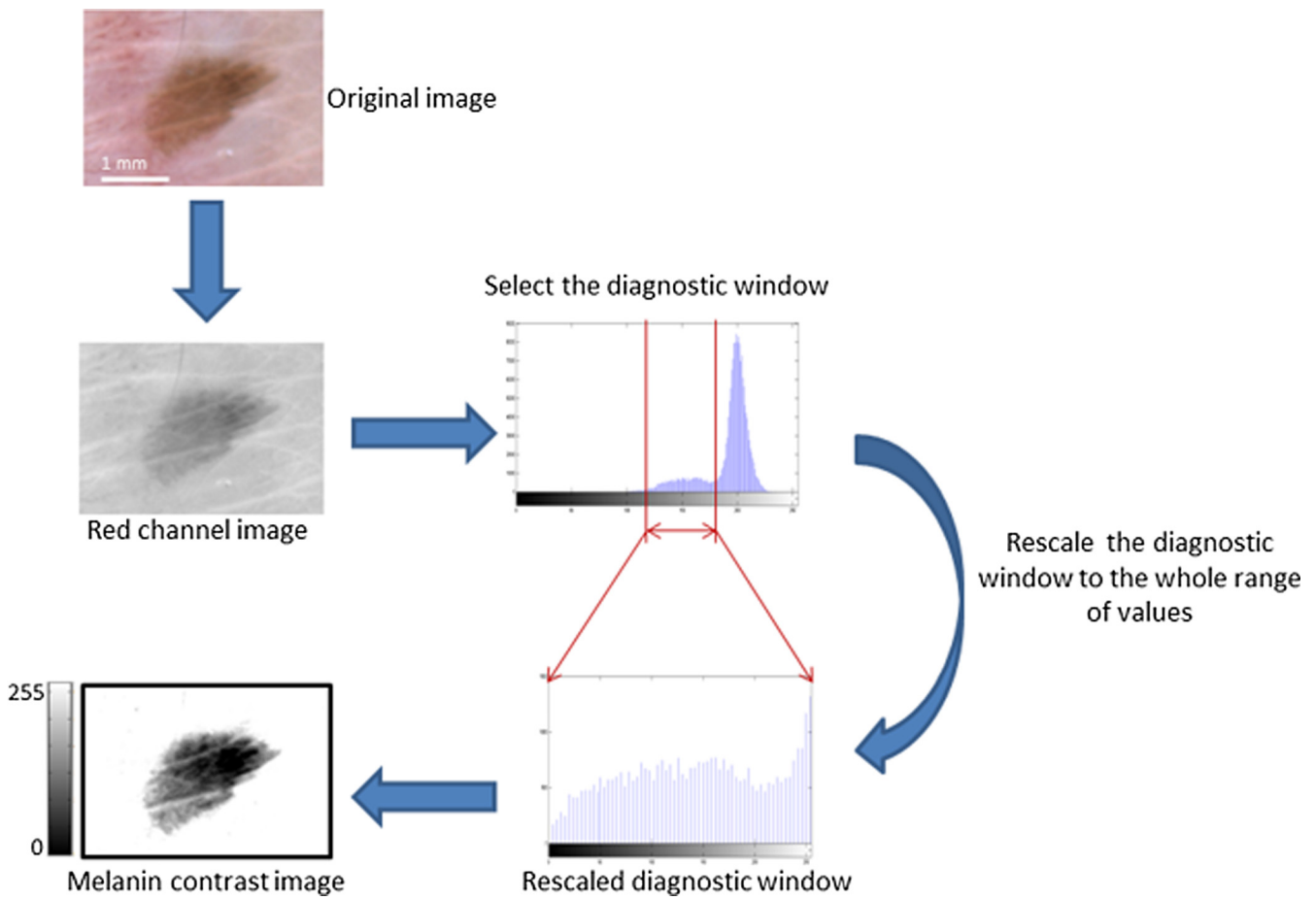

Fig. 6 Algorithm for the melanin contrast enhancement (MCE) mechanism on an image taken by our NCDD system.

for particular channels could also be improved from an otherwise "noisy" output.

4 Results and Discussion

The usefulness of our light source architecture is clearly evident in the first experiments with our dermoscope prototype, in the context of the aforementioned BCE and MCE mechanisms. Skin patches were illuminated in two different cases using the two remote phosphor modules described before. In Fig. 7, the image processing results for BCE are compared. As argued earlier, one can clearly see the advantage of having more red and green contents in the light source in order to have greater details on vascularity or blood contrast. After applying the BCE algorithm, a standard Gaussian lowpass filter command was used in both sets of images in order to minimize noise. Note that the multiplying factor mentioned in Fig. 5 was the same in both cases of the image processing results shown in Fig. 7. The variation of this multiplying factor alone during postprocessing cannot enhance the contrast beyond a certain limit, unless the red and green contents of the light source are suitably improved too. It is worth mentioning that the white balance setting in the camera software was kept in "auto mode" and our results clearly show that such a software adjustment is not always enough to compensate for spectral shortcomings of a light source in any imaging system. Hence, the spectral tuning of light sources can be very crucial for quality imaging. One prominent example is the "True Tone" dual-flash in Apple iPhone 5S.[14]

The same kind of comparison is shown in Fig. 8, where the MCE mechanism, without any manipulation of the original histogram, as shown in Fig. 6, was applied. In this case, a benign nevus with high melanin content in combination with blood vessels (see marked area in Fig. 8) was chosen. Therefore, the effect of hemoglobin contrast reduction by the use of a lower color temperature by the MCE mechanism is clearly visible. This makes it easier for physicians to distinguish between melanin and hemoglobin. The calculation of the dynamic range of the resulting contrast, via the Michelson contrast formula, with respect to the maximum concentration of melanin within the nevus and the surrounding skin area showed equally good values (melanin contrast value at 6703 K is 0.997 and at 3461 K, 0.987) for both color temperatures. However, the blood contrast could be reduced by about 25% compared to the contrast achieved with the higher color temperature (blood contrast value at 6703 K is 0.102 and at 3461 K is 0.075). By using even lower color temperatures, which is the goal in future studies, this effect could be further strengthened.

The theoretical background and the first results achieved are in good agreement. In cooperation with our medical partners from the University Medical Center Göttingen (UMG) and the Hannover Medical School (MHH), preclinical trials on more complex nevi samples from various individuals will be pursued in the next steps, using our novel light-source-based noncontact dermoscope.

It is worth mentioning that Hb-melanin crosstalk can affect the typical image enhancement techniques. An example is shown in Fig. 9, where the BCE technique has enhanced the melanin contrast unwarrantedly. So, in a vascular nevus with high melanin concentration, the BCE results could have unwanted artifacts due to the crosstalk with melanin contrast. However, in the MCE technique, blood contrast is suppressed, as shown in Fig. 8. Note that hyperspectral imaging with advanced algorithms has been shown to address this issue successfully.[5] Hence, this kind of

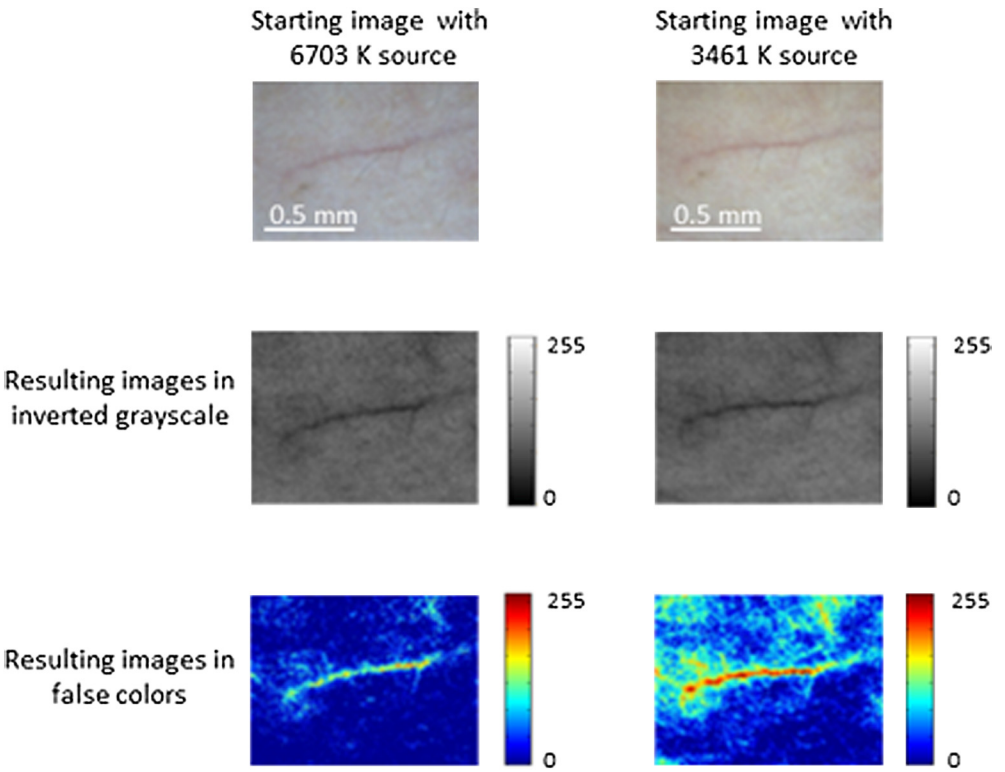

Fig. 7 Comparison of the phosphor performance for the BCE mechanism (left column: Ph1; right column: Ph2).

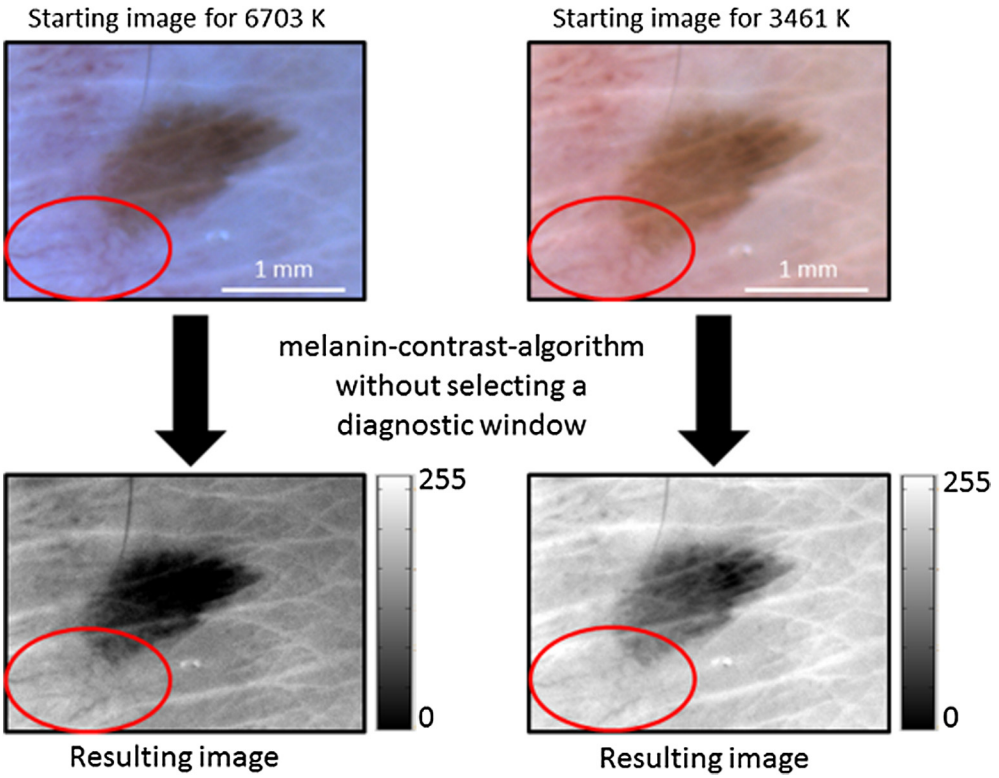

Fig. 8 Comparison of the phosphors for MCE mechanism (left column: Ph1; right column: Ph2).

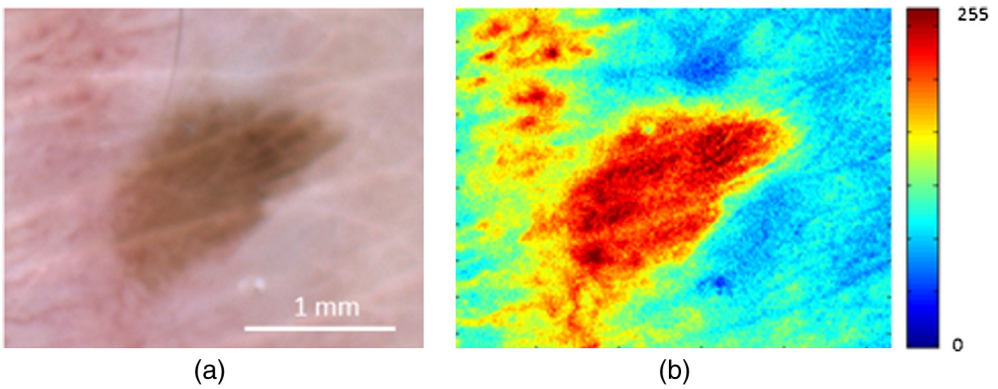

Fig. 9 Effect of crosstalk between melanin and hemoglobin in the BCE mechanism: (a) original image (as in Fig. 6) illuminated with Ph2; (b) resulting image after application of the BCE mechanism.

approach will be undertaken in our future experiments. Future experiments will also target the segregation of oxygenated-Hb and deoxygenated-Hb in a modified BCE approach.

Note that the focus of this article is the novel, cost-effective, and simple lighting architecture with the advantage of easy spectral tuning for better skin imaging. Keeping in mind the multidisciplinary nature of the light source, imaging, and image processing parts, we have chosen very simple image processing algorithms to explain the potential of the hardware system. Such simple image processing approach might suffer from Hb crosstalk, as mentioned in various publications. However, we expect that future work on customized and more sophisticated image processing tools to be used with our lighting system and imager can mitigate such issues.

5 Conclusion

A novel light source design, with a high-power blue LED pumped remote phosphor module, for an NCDD has been developed and experimentally verified. To the best of our knowledge, this is the first time the application of remote phosphor architecture has been demonstrated in a digital dermoscope, achieving easy and cost-effective spectral tuning. This concept can be used in a vast range of other biomedical imaging systems including microscopes. The extraordinary feature of this architecture is that the remote phosphor module is easily replaceable; hence, the output spectra can be customized depending on the CCT or spectral requirements, for skin imaging in particular to our case or any other kind of biomedical imaging in general. In applications where variation in the light spectra could be useful,

such remote phosphor architecture would possibly be economically more viable than a similar white LED architecture with fixed CCT and spectrum, for changing the white LED is more cost-intensive and often time consuming. Note that although two commercial remote phosphor modules were used in our prototype, in the future, many other kinds of phosphor fabrication recipes can be utilized to customize the spectra, in accordance with the needs of biomedical imaging. In the context of BCE and MCE mechanisms, the first results of our system are very promising. Further preclinical trials of our NCDD will follow, in order to study complex lesions (melanoma, psoriasis, acne, etc.) with different types of pigmentations and vascularity under customizable spectra from our remote phosphor-based light sources.

Acknowledgments

This work was supported by the Federal Ministry of Economics and Technology (BMWi) within the program Zentrales Innovationsprogramm Mittelstand (ZIM).

References

1. A. Günther et al., "An ultra-bright white LED based non-contact skin cancer imaging system with polarization control," *Proc. SPIE* **8798**, 87980O1 (2013).
2. M. Moncrieff et al., "Spectrophotometric intracutaneous analysis: a new technique for imaging pigmented skin lesions," *Br. J. Dermatol.* **146**(3), 448–457 (2002).
3. M. Elbaum et al., "Automatic differentiation of melanoma from melanocytic nevi with multispectral digital dermoscopy: a feasibility study," *J. Am. Acad. Dermatol.* **44**(2), 207–218 (2001).
4. D. Kapsokalyvas et al., "Spectral morphological analysis of skin lesions with a polarization multispectral dermoscope," *Opt. Express* **21**(4), 4826–4840 (2013).
5. F. Vasefi et al., "Polarization-sensitive hyperspectral imaging *in vivo*: a multimode dermoscope for skin analysis," *Sci. Rep.* **4**, 4924 (2014).
6. N. Narendran, "Improved performance white LED," *Proc. SPIE* **5941**, 45–50 (2005).
7. http://www.lumaera.com/wp-content/uploads/2013/04/LUMAERA-50-RP.pdf (last accessed on May 22 2015).
8. C. Basu, M. Meinhardt-Wollweber, and B. Roth, "Lighting with laser diodes," *Adv. Opt. Technol.* **2**(4), 313–321 (2013).
9. V. J. Nadeau et al., "Laser-pumped endoscopic illumination source," in *Conf. Proc. IEEE Engineering in Medicine and Biology Society*, Vol. 2008, pp. 2059–2062 (2008).
10. http://www.luminus.com/products/Luminus_CBT90_Datasheet.pdf (last accessed on May 22 2015).
11. F. Fanjul-Vélez et al., "Determination of the pathological state of skin samples by optical polarimetry parameters," *Proc. SPIE* **7138**, 71380I (2008).
12. G. Zonios et al., "Melanin absorption spectroscopy: new method for noninvasive skin investigation and melanoma detection," *J. Biomed. Opt.* **13**(1), 014017 (2008).
13. D. Kapsokalyvas et al., "Multispectral dermoscope," *Proc. SPIE* **7368**, 73680D (2009).
14. https://www.apple.com/iphone-5s/camera/ (last accessed on May 22 2015).

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Bernhard Roth obtained his PhD in physics at Bielefeld University in 2001. From 2002 to 2007, he was research group leader at Duesseldorf University and obtained his state doctorate (Habilitation) in quantum optics in 2007. Since 2012, he has been director of the Hannover Centre for Optical Technologies and since 2014, professor of physics at Hannover University. His scientific activities include laser spectroscopy, polymer optical sensing and optical technology for illumination, information technology, and the life sciences.