Skin biomedical optical imaging system using dual wavelength polarimetric control with liquid crystals

Avner Safrani¹, Ofir Aharon¹, Shahar Mor¹, Ofer Arnon², Lior Rosenberg²

and I. Abdulhalim¹

¹Department of Electro-Optic Engineering, Ben Gurion University of the Negev, Beer Sheva 84105, Israel

²Department of Plastic and Reconstructive Surgery and Burn Unit, Soroka University Medical Center, Ben Gurion University of the Negev, Beer Sheva 84105, Israel

Spectropolarimetric skin imaging is becoming an attractive technique for early detection of skin cancer. Using two liquid crystal retarders in combination with dual band passive spectral filter and two linear polarizers we demonstrate the spectral and polarimetric imaging of skin tissue in the near infrared. Based on this concept a compact prototype module is built and being used for clinical evaluation.

**Keywords:** skin imaging, skin cancer, spectral imaging, polarimetric imaging.
Introduction

The skin is the outer coverage of the body, and its largest organ. The skin covers and protects the underlying muscles, bones, ligaments and internal organs. The main functions of the skin are to provide sensation, insulation and temperature regulation. The skin protects the body from water loss, pathogens exposure, injuries and chemical insults. The skin also provide protection from ultra violate radiation. The skin may be damaged by external and internal insults. For example: burns, injuries and tumors. The process in which the skin repairs the damage, called "wound healing" which ends by the epithelialization of preserved dermis or formation of a scar. During the phase of wound healing the border between healthy and damaged skin is difficult or even impossible to define due to the inflammatory and regenerative process. The significance in determining healthy borders is important in tumor treatments (excisions). Other aspect of interface between two structures in healthy skin is the border of a mole or other cutaneous lesions. The shape of the borders of a mole is one of the key factors in the clinical diagnosis of skin cancer. A malignant transformation may be assessed non invasively by correctly defining the mole's borders. It is important to be able to assess both the superficial layer and deeper layers of the skin. Four parameters indicate malignancy potential (ABCD) two of them A & B, Asymmetry and Borders, define the lesion interface with the normal tissues. Sun exposed skin is prone to develop non pigmented lesions from the group of Basal Cell Carcinoma (BCC- as many as 75% of all the skin cancer cases) and Squamous Cell Carcinoma (SCC) as many as 20%. The sun exposed face and neck are the most prone to develop these lesions [1]. The tendency to limit excision to minimum because of cosmetic considerations asks for accurate definition of the borders (B) of these tumors. In melanoma skin cancer, the most deadly skin cancer type, the importance of
non invasive, fast early detection is much more significant. Melanoma is an irregularly proliferating pigmented melanocyte cells. If not treated in early stages, melanoma can metastasize to vital organs and eventually cause death. At present, the currently used diagnostic technique for Melanoma detection is the expert's visual assessment, sometimes with the assistance of a dermatoscope. At present, definite diagnosis can be achieved only by invasive tissue biopsy or excisional biopsy. Non invasive technique for accurate early detection of skin cancer and its boundaries could be an important tool in this field.

During the last couple of decades several noninvasive methods were suggested, among them methods based on light scattering [2-3], confocal microscopy[4], photo acoustic microscopy [5], polarimetric imaging [6-8] and spectropolarimetric (SP) imaging [1,9,10]. In the SP imaging method, images are acquired at different wavelengths and different linear polarization states. Usually two polarized images are used, one with parallel polarizer and analyzer and the second with crossed polarizers. The subtraction of the perpendicular polarization image from the parallel polarization image provides an image which is mainly composed of the superficial layer of the skin (0.1-0.3 mm.), whereas the subtraction of the spectrally resolved images provides an image of deeper skin layers (down to 1 mm) [9]. The SP method holds several advantages with respect to other methods. With respect to the confocal methods it is much faster and cheaper; with regards to scattering methods it is cheaper, faster and provides a visualization of the lesion; with regards to the polarimetric-only method it has an additional degree of freedom by providing spectrally resolved images and thus deeper skin layer images with improved contrast due to different absorption and scattering characteristics of benign and malignant tissues. In SP it is important to be able to scan the wavelength and polarization fast
enough both for the convenience of the patient and doctor and to prevent the blurring effects of unintentional movements. Compactness and miniaturization are important factors for the ease of use. Recently we have developed several liquid crystal (LC) tunable filters [11-12] and polarization control [13] devices for the purpose of using them in biomedical optical imaging systems. In this study we present one implementation of these concepts by which we acquire polarimetric images at two different wavelengths controlled with simple LC retarders in the near infrared (NIR).

2. Liquid Crystal NIR Dual Wavelength Polarization Controller

The following spectropolarimetric method uses an LC device that generates different wavelength polarization modes. We present the theoretical bases, as well as the practical performances of the diagnostic LC device.

2.1 Theoretical design

The device's optical configuration is depicted in Fig.1. The light entering the device is linearly polarized along the x axis direction by the first polarizer (P1), followed by a liquid crystal retarder (LCR1) which is aligned with its optical axis at 45 degrees with respect to the y axis. The light is then transmitted towards the second polarizer (P2) which is parallel to the first polarizer. An additional LC retarder LCR2 which is parallel to LCR1 is positioned after the second polarizer. Finally, a dielectric mirror (Di), an optical long pass filter (L) and an optical short pass filter (S) are positioned at the end of the optical stack.
Fig.1. The optical stack comprising the LC device for the wavelength and polarization control.

The output field that is transmitted through the optical components in Fig.1 is given by the Jones matrix multiplications as follow:

\[
\vec{E}_{\text{OUT}} = S(\omega_{\text{HPF}}) \cdot L(\omega_{\text{LPF}}) \cdot Dl(\Delta\omega) \cdot \text{LCR2}(\delta_2, 45^\circ) \\
\cdot P2(0^\circ) \cdot \text{LCR1}(\delta_1, 45^\circ) \cdot P1(0^\circ) \vec{E}_{\text{IN},0} \exp(i\omega t)
\]

(1)

Where \( \omega_{\text{HPF}} \) and \( \omega_{\text{LPF}} \) are the cutoff frequencies of the short pass filter and long pass filter, respectively, and \( \Delta\omega = \omega_2 - \omega_1 \) is the forbidden frequency band of the dielectric mirror. The retardation of the LCRs is denoted as \( \delta (\delta = 2\pi d(n_e - n_o)/\lambda) \) and it is controlled by voltage. Thus, the Jones matrices formalism [14] of eq.1 is:

\[
\vec{E}_{\text{OUT}} = \left[ \begin{array}{cc}
  u(\omega - \omega_{\text{HPF}}) & 0 \\
  0 & u(\omega - \omega_{\text{LPF}})
\end{array} \right] \left[ \begin{array}{cc}
  1 - u(\omega - \omega_{\text{HPF}}) & 0 \\
  0 & 1 - u(\omega - \omega_{\text{LPF}})
\end{array} \right] \\
\times \left[ \begin{array}{cc}
  1 - u(\omega - \omega_1) + u(\omega - \omega_2) & 0 \\
  0 & 1 - u(\omega - \omega_1) + u(\omega - \omega_2)
\end{array} \right] \\
\times \exp(i\beta_2) \left[ \begin{array}{cc}
  \cos \delta_2 / 2 & -i \sin \delta_2 / 2 \\
  -i \sin \delta_2 / 2 & \cos \delta_2 / 2
\end{array} \right] \\
\times \left[ \begin{array}{cc}
  1 & 0 \\
  0 & 0
\end{array} \right] \exp(i\beta_1) \left[ \begin{array}{cc}
  \cos \delta_1 / 2 & -i \sin \delta_1 / 2 \\
  -i \sin \delta_1 / 2 & \cos \delta_1 / 2
\end{array} \right] \left[ \begin{array}{cc}
  1 & 0 \\
  0 & 0
\end{array} \right] \vec{E}_{\text{IN},0} \exp(i\omega t)
\]

(2)
Where $\beta_{1,2}$ are the average phase accumulations in passing through the retarders and the function $u$ is a unit step function,

$$u(x) = \begin{cases} 1 & x \geq 0 \\ 0 & \text{else} \end{cases} \quad (3)$$

By performing the multiplication, the resulting electric field is given by equation 4:

$$\tilde{E}_{\text{OUT}} = u(\omega - \omega_{\text{HPF}})[1 - u(\omega - \omega_{\text{LPF}})]\left[1 - u(\omega - \omega_1) + u(\omega - \omega_2)\right]$$

$$\times \left[ \frac{\cos \delta_2 / 2 \cos \delta_1 / 2}{-i \sin \delta_2 / 2 \cos \delta_1 / 2} \right] E_{\text{IN},0,X} \exp(i(\omega t + \beta_1 + \beta_2)) \quad \text{(4)}$$

Hence, the output electric field has both vertical and horizontal components and its amplitude is confined within the band pass filter (BPF) none zero values created by the $u$ functions, see Fig.2 (for simplicity the frequency dependence was replaced with the wavelength):

Fig.2. Qualitative output of the bandpass filter created by the combination of an HPF, an LPF and a suitable dielectric mirror

Since the LCR retardation is electrically controlled, the optical device may perform as linear polarization switch by changing the retardation of the second LCR from $\delta_2 = 2\pi m$ to $\delta_2 = (2m+1)\pi$ - the resulting electric fields are given in eq.5 and eq.6, respectively (in terms of the wavelength):
The electric field in equation 5 is parallel to the x axis, whereas the electric field of equation 6 is perpendicular to the x axis. Moreover, if the first LCR is designed in accordance with the band pass filter transmission, it is possible to choose either one of the allowed bands from the BPF by controlling the retardation of the first LCR. By changing the supplied voltage for the first LCR one may set \( \frac{2}{\cos \delta} \) to be zero for the first peak of the BPF, and \( \frac{2}{\cos \delta} \) to be a maximum for the wavelength at the second peak of the BPF. Then, by controlling the retardation of the second LCR, linear polarization switching may be performed. Alternatively, the first peak of the BPF is transmitted and the second peak is blocked and yet again linear polarization switching may be performed. The above statements are clarified by writing the transmission expression for the parallel and perpendicular polarization states:

\[
\tilde{E}_{\text{OUT}}|_{\delta_2=2,2m} = u(\lambda - \lambda_{\text{HPF}})[1-u(\lambda - \lambda_{\text{LPF}})][1-u(\lambda - \lambda_i)+u(\lambda - \lambda_2)] \\
\times \left[ \cos \frac{\delta_i}{2} \right] E_{\text{IN},0,X} \exp \left( i \left( \frac{2\pi \lambda}{\lambda} t + \beta_1 + \beta_2 \right) \right)
\]

\[
\tilde{E}_{\text{OUT}}|_{\delta_2=(2m+1)\pi} = u(\lambda - \lambda_{\text{HPF}})[1-u(\lambda - \lambda_{\text{LPF}})][1-u(\lambda - \lambda_i)+u(\lambda - \lambda_2)] \\
\times \left[ -i \cos \frac{\delta_i}{2} \right] E_{\text{IN},0,X} \exp \left( i \left( \frac{2\pi \lambda}{\lambda} t + \beta_1 + \beta_2 \right) \right)
\]

\[
I_{\text{par}} = I_{\text{per}} \propto \left| \tilde{E}_{\text{OUT}} \right|^2 \\
= u(\lambda - \lambda_{\text{HPF}})[1-u(\lambda - \lambda_{\text{LPF}})][1-u(\lambda - \lambda_i)+u(\lambda - \lambda_2)]^2 \\
\times \cos^2 \frac{\delta_i}{2} : E_{\text{IN},0,X}^2
\]

\[
= U(\lambda)E_{\text{IN},0,X}^2 \cos^2 \frac{\delta_i}{2}
\]

\( \lambda \), \( \lambda_i \), \( \lambda_{\text{HPF}} \), \( \lambda_{\text{LPF}} \), \( \beta_1 \), \( \beta_2 \), \( \delta_i \), are the appropriate parameters for these expressions.
2.2 Device Practical Characterization

The experimental setup is depicted in Fig.3. A fiber wide spectral source (Dolan-Jenner DC-950H DC-Regulated Fiber Optic Illuminator) was used to illuminate the device (the NIR constant filter of the illuminator was extracted from the apparatus). The emerging light was collimated by an achromatic objective lens (O). A first NIR polarizer (TECHSPEC NIR Linear Polarizer, extinction ration 40 dB, 750-850 nm) was positioned vertically (P1) followed by a first LCR at 45° orientation (LCR1) and a second vertically aligned NIR linear polarizer (P2). A second LCR at 45° orientation (LCR2) is positioned after the second linear polarizer followed by a long pass (L) filter ($\lambda>700$ nm, Edmunds optics, T > 97%), a short pass (S) filter ($\lambda<950$ nm, Edmunds optics, T > 60%) and a dielectric (Di) mirror (CVI Optical Components and Assemblies, TMI1 800, R > 99%). The LC (Merck E44) retarders were constructed in our lab (clean room, grade 100). The first LCR was constructed with silica spacers with a diameter of 10 microns whereas the second LCR spacers were 4.7 microns in diameter. The entire stack was followed by an additional linear polarizer, A, (polarization plate beam splitter, Edmunds Optics), an achromatic collecting lens (C), a collecting optical fiber supplied with a fiber connector (FC) and a parallel spectrometer (StellarNet Inc.). The LCR's were supplied with 1kHz sinusoidal wave with suitable voltages.

Fig.3 The optical setup used for the characterization of the LC dual wavelength polarization controller
The absolute transmission of the device and the polarization linearity were tested. In Fig.4 the optical normalized transmission, as well as the extinction ratio, of the device are depicted for both vertical and horizontal polarization. In Figs.4a-4b the transmission of the 729 nm and 922 nm vertical and horizontal polarization modes of the device are depicted, respectively. The obtained signals are strong enough for our imaging system and have a relatively narrow band (FWHM < 25 nm). In Figs.4c-4d the extinction ratio was measured for both the vertical and the horizontal polarizations of the 729 nm and the 922 nm lines, respectively. The extinction ratio for the 729 nm line is 35 for the vertical polarization and 250 for the horizontal polarization. The extinction ratio for the 922 nm line is 10 for the vertical polarization and 72 for the horizontal polarization. Note that there is a small none ideality both in the polarization linearity and in the filtering strength. However, these two factors may be further improved by an optimization process of the LCR design and construction.
Fig. 4 The spectropolarimetric performances of the device for the 729 nm and 922 nm lines. (a) and (b): vertical and horizontal normalized transmission of the 729 nm and 922 nm lines, respectively. (c) and (d): vertical and horizontal extinction of the 729 nm and 922 nm lines.

3. Spectropolarimetric Imaging System

The spectropolarimetric imaging system is depicted in Fig. 5. A fiber bundle consisting of 50 micron core diameter optical fibers transmitting in the spectral range 400 nm - 2000 nm is used as the light guide in the optical system (Dual Branch Light Guide, Edmunds Optics). The optical fiber connector (FC) is concentric with the objective entrance pupil. The light emerging the objective is collimated and then linearly polarized by the first NIR linear polarizer (P1). After the first linear polarizer, a first liquid crystal retarder (LCR1) is positioned followed by a second NIR linear
polarizer (P2). LCR1 is used to pick one of the two spectral lines 729 nm or 922 nm, whereas LCR2, which is positioned right after the second linear polarizer, is used to control the direction of the transmitted linear polarization (vertical or horizontal). A metal scanning-plain-mount (SPM) is positioned at the object plain. The SPM has a circular window where the diagnosed skin is to be positioned. The back scattering of the object is then reflected towards an additional linear polarizer (A) which is vertically aligned, and then collected by an achromatic lens (C). A long pass filter (L), a short pass filter (S) and a dielectric mirror (Di) are positioned between the collection lens and the NIR CCD camera (Sony XC-E Series Monochrome CCD Camera, spectral range 400-870 nm). The setup can photograph lesions of 15 mm x 12 mm in size and the angle between the illuminating optical path and the imaging optical path is 40 degrees, although any angle between 10 to 50 degrees was checked to give similar results.

Fig. 5 The spectropolarimetric imaging system on an optical bench
The SP imaging system is automatically controlled by MATLAB. Four different images, in less than half a second, are taken for each lesion, two for each spectral line in both horizontal and perpendicular polarizations. After acquiring the images, a simple image processing algorithm is used to enhance the quality of the image and to obtain the superficial and deep skin layers images. The method of isolating the backscattering photons reflected by superficial skin tissue from the deeply diffused (into the dermis) dominant photons is done by subtracting the horizontal polarization image (which is the image where the impinging polarization is at crossed state with respect to the analyzer polarization orientation) from the vertical polarization image. The method assumes that linearly polarized light impinging on skin tissue is depolarized by the superficial skin tissue and the deeper the light penetrates the greater is the depolarization effect. Hence, photons reflecting backwards from deep skin layers are totally depolarized, whereas photons reflecting from the more superficial skin layers maintain some certain degree of polarization. By subtracting the horizontal polarization image from the vertical one we recover the photons that are backscattered from the superficial layer only. The method improves the quality of the surface image and hence allows a better diagnosis of the nevi nature: (symmetry and borders). The above process is done for each wavelength and since the scattering and penetration nature of the photons at different wavelengths are different the process produces different depth images with great borders appearance. In addition, by subtracting longer wavelength images from shorter wavelength images (i.e. 729 nm images subtracted from 922 nm image) deep tissue images are revealed allowing assessment of lesion's deeper structures. To avoid the differences between the illumination intensity of the spectrally resolved images, before subtraction of the
wavelength images, a normalization process is applied to each image. To reduce the small spatial noise, the result of small movements of the subject while photographing (in spite the speed of 0.1 second that the images are taken) we convolved all the images with a spatial Gaussian filter and thereby reducing the influence of any random noise as well as small pixel size movements of the subject. The Gaussian filter had a mean and variance values of 0 and 2, respectively and the convolving matrix included 11x11 pixels.

4. Skin imaging results with an optical bench setup

The above imaging method was applied to few different lesions using optical bench setup shown in figure 5 before building a suitable prototype for clinical tests. In Fig.6 a small scar positioned on the palm of the hand was photographed. The image of Fig.6a is much sharper than the image of Fig.6b. The photons of the image of Fig.6b experienced much more scattering events carrying tissue information from deeper skin layer; hence they are more depolarized by the many scattering events. Note that the deformed scar tissue is seen under the superficial epidermal layer. In Fig.6a the superficial layer of the skin is visualized. The photons of the image of Fig.6a are photons that maintain their polarization since they are back scattered from the superficial layer of the skin. Fig.6c and 6d are similar images to those in Fig.6a and 6d except for the different wavelength. Note that there are resolution differences between the wavelength images of the 729 nm and the 922 nm, which occurs because the penetration depth of the 922 nm photons is greater than that for the 729 nm photons so that different tissue depth images are obtained, therefore the images of the 922 nm are more blurred. In Fig.7 the scare images after the simple image processing are depicted. In Fig.7a the difference between the image at 922 nm and that at the 729
nm with vertical polarization is presented. Note that it highlights the bottom of the scar. The image at Fig.7b is too blurred, apparently because the horizontal images of both the 729 nm and the 922 nm are too blurred. However, the polarimetric images at 729 nm and 922 nm are with high resolution and many details are exposed. Note that in each one of the images (apart from the image in Fig.7b) the borders and the shape of the scar are highly defined.

Fig. 6 Vertical and Horizontal images of a scar located on the palm of white Caucasian: (a) vertical image at 729 nm, (b) horizontal image at 729 nm, (c) vertical image at 922 nm, (d) horizontal image at 922 nm.
Fig. 7 Polarimetric and spectral image processing: (a) spectral vertical difference image, (b) spectral horizontal difference image, (c) polarimetric image at 729 nm, (d) polarimetric image at 922 nm
Fig. 8 Vertical and horizontal images of a Seborrheic Keratosis treated by Cryosurgery (tissue destruction by freezing): (a) vertical image at 729 nm, (b) horizontal image at 729 nm (c) vertical image at 922 nm, (d) horizontal image at 922 nm

In Fig. 8 images of a 3 mm Seborrheic Keratosis on the dorsal side of the palm of a white Caucasian subject is, treated by cryosurgery. Note (Fig. 8a and 8c) that the contour of the lesion is well defined. Fig. 8 and figure 6 represent two different subjects under identical technical imaging conditions. Fig. 9 is same image of Fig. 8 after applying the simple image processing algorithm. Note that in Fig. 9a it seems that the deeper part of the lesion is represented with the contours easily recognized. In Fig. 9b the contour of the lesion is also seen and in Fig. 9c and 9d the edges of the lesion, in two different tissue depths, are furthered enhanced. We examined the lesion after the frozen tissue was removed and found that indeed the shape of the cavity, was
very similar to the image of Fig.9d. The thickness of the frozen tissue was over a one millimeter which we estimate to be roughly the optical penetration capability.

Fig.9 Polarimetric and spectral image processing of a Seborrheic Keratosis treated by Cryosurgery: (a) spectral vertical difference image, (b) spectral horizontal difference image, (c) polarimetric image at 729 nm, (d) polarimetric image at 922 nm.
Fig. 10 Vertical and horizontal images of a non-treated Seborrheic Keratosis: (a) vertical image at 729 nm, (b) horizontal image at 729 nm, (c) vertical image at 922 nm, (d) horizontal image at 922 nm.

In Fig. 10, images are shown of non-treated Seborrheic Keratosis, 2.5 mm in diameter which is located on the hand of a white Caucasian subject. The lesion protrudes about a half of millimeter above the superficial layer of the skin. Note that the lesion is not seen in Fig. 10d apparently because the imaged tissue is deeper than the lesion borders. Fig. 11 is the image of Fig. 10 after applying the simple image processing. Note that in Fig. 11a it seems as if the lesion's deep border has been demonstrated and that the contour of the lesion in the deep tissue is easily recognized. In Fig. 11b apparently a more superficial tissue level was demonstrated imaged and here as well the contour of the lesion is easily seen. In Fig. 11c the boundaries of the
The superficial layer of the lesion are further enhanced. The image reveals the precise borders of the lesion along with its spatial structure – it is easy to recognize the protrusion of the lesion above the surrounding skin.

![Image](image-url)

Fig.11 Polarimetric and spectral image processing of a non-treated Seborrheic Keratosis: (a) Spectral vertical difference image, (b) spectral horizontal difference image, (c) polarimetric image at 729 nm, (d) polarimetric image at 922 nm.

5. Skin imaging results with clinical prototype

The experimental setup was assembled into a useful compact clinical module. The module is presented in Fig.12. It is constructed of two movable arms in a tilted V-shape which comprises the optical components as described in the experimental setup. The body of the prototype module was fabricated using an STL printer (Stereo...
Lithography Printer of Objet). The angle between the arms is 25 degrees and the total length of the device is 30 cm. Due to its shape we simply called the device V-imager. The V-imager comprises a circular window at the object plan with a diameter of 23 mm allowing imaging and assessment of such size lesions. The V-imager was used in the Department of Plastic Surgery at Soroka University Hospital in Beer Sheva. An example of a characteristic image of a compound nevus of a white Caucasian subject is depicted in Fig.13.

![V-imager clinical prototype](image)

**Fig. 12 The V-imager clinical prototype**

Fig.13 is identical to Fig.11 in all aspects except for the imaged object. Note that the contours of the lesion as well as some pigmentation differences are easily seen. Naked eye observation of this lesion does not reveal any pigmentation differences and it is impossible to assess its distribution beneath the skin superficial layer. To obtain
the image in Fig.14, first the vertical polarization image at 729 nm is subtracted from
the vertical polarization image at 922 nm to create the vertical polarization
wavelength difference image. Second, the horizontal polarization image at 729 nm is
subtracted from the horizontal image at 922 nm to create the horizontal polarization
wavelength difference image. Then to achieve the image of Fig. 14, the horizontal
polarization wavelength difference image is subtracted from the vertical polarization
wavelength difference image.

Fig.13 Vertical and Horizontal images of a compound nevus of a white Caucasian
subject: (a) vertical image at 729 nm, (b) horizontal image at 729 nm, (c) vertical
image at 922 nm, (d) horizontal image at 922 nm.

In Fig.14 it seems as if the nevus's deep border has been demonstrated and its
contour in the depth of cutaneous tissue is easily recognizable. By subtraction of the
vertical polarization of the spectrally resolved images we obtain an image of deep skin layer which is not depolarized. By subtraction of the horizontal polarization of the spectrally resolved images we obtain an image of deep skin layer which is grossly depolarized. By subtracting these two images we eliminate the backscattering from the vertical difference image, thus obtaining a sharp image of the contour of the object underneath the superficial layer.

Fig.14 Polarimetric and spectral difference image
6. Conclusions

A novel concept is proposed for spectropolarimetric imaging using specially designed LC devices incorporated into miniature imaging module used for skin imaging. The LC device characteristics matched the theoretical design. It should be pointed that by using additional dielectric mirrors, for example dielectric mirrors with central wavelengths of 500 nm and 700 nm, along with a shorter pass filter (for example a short pass filter of $\lambda>$400 nm and a long pass filter $\lambda<$800) more spectrally resolved images can be obtained. In this respect, a 400 nm spectrally resolved image will provide additional information since melanoma skin cancer cells tend to absorb strongly this wavelength. The implementation of the SP system along with a simple image processing algorithm produced SP images set which can help clinicians to differentiate the precise borders and shape of a lesion on the surface and deeper below the epidermal tissue. The system was tried on a small scar, an untreated and treated Seborrheic keratosis and a compound nevus of a white Caucasian subjects. The result showed clear images of the shape and borders of the superficial layer of the lesions along with, what we believe to be, deep into the epidermal tissue. A clinical prototype was designed, built and used to obtain more SP images of patients in the hospital and to collect more data which eventually may allow diagnosing skin cancer and in particular melanoma skin cancer in early stages. The system is unique in several aspects: it is fast, compact, not costly, it has no mechanical moving parts, it uses a halogen lamp illumination which allows a wide spectrum along with true color imaging, it applies NIR spectra and thus has penetration capabilities deep bellow the epidermis.
References


