Quasi-holographic solution to polarization-sensitive optical coherence tomography acceptable to nonlaboratory applications

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Abstract. Experimental proof-of-concept is presented for a quasi-holographic solution to polarization-sensitive optical coherence tomography (PS OCT). Due to decoupling between the reference and sample beams by polarization, the solution seems acceptable to acquisition and communication of optical data in the nonlaboratory environment. The nonlab environment implies uncontrollable disturbances, e.g., temperature changes and mechanical effects happening under shop testing in industry or routine examinations in common clinics and hospitals. For mapping the collagen-related depolarization ratio of light backscattered from the human dermis, a phenomenological model is evolved from the theory of light depolarization in crystalline polymers. The model yielded a simplified intensity-based estimation algorithm. The design concept and the model rely on a submillimeter tumor thickness as a proofed prognostic factor and an important criterion for complementary functional diagnostics of skin cancers in their early phase. Choice of the model is inspired by similarity of structural and optical properties between liquid-crystal collagen fibers in the dermis and birefringent crystalline lamellae in some polymer materials. The model gives a plausible interpretation of a peculiarity of cumulative birefringence in the abnormal skin dermis. Following a top-down approach to design, the authors attempt to contribute to bridging the gap between practitioners' concerns and academic studies, © 2008 Society of Photo-Optical Instrumentation Engineers. [DOI: 10.1117/1.2967996]

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1 Introduction

It has been more than 30 years since A.P. Ivanov, A.P. Chaikovskii, and A.A. Kumeisha advanced and proved experimentally a new method for optical sectioning in scattering, not only reflecting, media by means of depth-resolved measurements of backscattered light with low coherence interferometry. A potential for the study of the microstructure of blood, living tissue or leaf, etc. was directly asserted in their pioneer paper. This method is the immediate precursor to optical coherence tomography (OCT). It should be evident to familiar with the literature. Recently, the above-mentioned paper has been mentioned by de Boer in his comprehensive reference.

Nowadays, the penetration depth for cross-sectional imaging performed with OCT is limited to a little more than one millimeter for scattering biological tissues such as the human skin. At first glance, it could impede the application of OCT. But biomedical imaging is too loose a concept. That is why a distinction should be made between biological subcellular studies and common clinical examinations. Smart optical techniques of subcellular resolution including OCT are able to successfully complement conventional biopsy and histopathology of superficial skin but are not readily available for routine clinical inspections. These techniques can identify images of cancer cells to only a few hundred micrometers depth. However, the recent detailed studies of cancer statistics in Germany, Austria, Switzerland, North America, Scotland, and Australia show that the tumor thickness of 0.5 mm dominates as the strongest prognostic factor of superior long-term survival for patients with thin cutaneous melanoma. Although the occurrence of melanoma is small, e.g., in comparison to basal cell carcinoma—one of the commonest types of malignant superficial lesions, often melanoma-type cancers are not diagnosed until they are in the later stage of develop-
Growing vertically through the epidermal-dermal junction, the tumor damages optically birefringent collagen fibers, an abundant structural and functional component of dermis. A depletion of collagen content manifests through a loss of cumulative (macro-) birefringence within the dermis regions invaded by tumor. Estimation of inferior and lateral tumor margins is essential to minimize the number of numerous histological excisions and preserve as much healthy tissue as possible. In general, the potential of polarization-sensitive OCT (PS OCT) to provide functional characterization of abnormal dermis by mapping these margins was demonstrated under laboratory conditions. However, for applications with unexpected external perturbations, e.g., mechanical effects and the ambient temperature drift arising in the clinical and hospital environment, PS OCT systems must meet special requirements. Before now, we made just speculative attacks on this challenge. Here, we reveal the basic principle and consider in more detail the experimental verification of the specific solution proposed for such systems as well as mention the simplified algorithm for mapping birefringence of dermis through measurements of the depolarization ratio.

2 Design Criteria

2.1 Top-Down Approach

 Needless to say, dermatologists and patients would prefer to deal with a skin probing head that looks like a friendly handheld device coupled to a flexible interface cable. The examinations should be free of tedious readjustments after each test or changes in ambient properties necessarily happening in common cancer care units or consulting rooms. It is also strongly desirable to display results of tests in more indicative form accessible to an ordinary medical operator. But the issues do not add up to ergonomic engineering only. As applied to PS OCT, the main problem is how to reduce deterioration of polarization data integrity in the course of acquisition and communication of optical information from a patient’s body to the operator workstation. The adequate solution is seen in meeting simultaneously the following basic requirements:

• Decoupling between the reference and sample beams by polarization to provide data transmission through separate polarization-preserving optical fibers;
• Avoiding any mechanical modulation in the reference arm as a weak point in the case of external mechanical impacts;
• Use of a simplified criterion for characterizing the margins of dermis regions with abnormal changes in collagen content.

Such a top-down approach has brought us to the solution called, for short, quasi-holographic. The solution gives instantaneous A-scans as spatially evolved fringes similar to those obtained in holographic spectroscopy. Since spectral content is beyond our interest here the fringe backscattering from the near interface i and the rear interface j. Let i-train and j-train spend times $T_{iMZ}$ and $T_{jMZ}$, respectively, for the total travel of the sample arm. Therefore, the concurrent reference r-train needs time $T_{rMZ}$ to pass through the reference arm. The length of this arm is adjusted so that

$$T_{iMZ} < T_{jMZ} < T_{rMZ}. \tag{1}$$

As seen in Fig.1, the r-, i-, and j-trains combine in PBS and travel BE to enter WP. For clarity, these trains are denoted in Fig.1 (lower right) as wave forms with orthogonal polarizations ⊗ and ⊕. Shown at the entry of WP, relative group delays between the wave forms are arranged according to the relationship.
Now, observe how the trains pass through WP—the basic element of WPI. The standard WP shown in Fig.1 comprises two \( \theta \)-angled wedges of equal middle thickness \( d \) jointed by their hypotenuses. The wedges are made of uniaxial negative crystal, preferably calcite, with birefringence \( n_e - n_o < 0 \), where \( n_e \) and \( n_o \) are extraordinary and ordinary refractive indices, respectively. As seen in Fig.1 (lower right) the optical axes \( \xi \) and \( \omega \) in the wedges are aligned perpendicular to each other and parallel to the entrance-exit faces of WP. Each of the combinations \((\xi, \omega)\) and \((\xi, \xi)\) is related to the retarded traveling with the speed \( c/n_e \), whereas \((\xi, \omega)\) or \((\xi, \xi)\) corresponds to the less retarded propagation with the speed \( c/n_o \).

For sufficiently small angle \( \theta \) and angles of incidence to the entrance face of WP, the following expression well approximates the time \( T_{j WP} \) needed for the \( j \)-train to pass through WP at the distance \( x_j \),

\[
T_{j WP} = \frac{n_e(d + x_j \tan \theta)}{c} + n_o(d - x_j \tan \theta),
\]

(2)

and time \( T_{r,i WP} \) of the \( r \)-train traveling WP at the same distance \( x_j \) is

\[
T_{r,i WP} = \frac{n_e(d + x_j \tan \theta)}{c} + n_o(d - x_j \tan \theta).
\]

(3)

Similarly, time \( T_{i WP} \) of the \( i \)-train traveling WP at the distance \( x_i \) is approximated as

\[
T_{i WP} = \frac{n_e(d - x_i \tan \theta)}{c} + n_o(d + x_i \tan \theta),
\]

(4)

and time \( T_{r,i WP} \) of the \( r \)-train traveling WP at the distance \( x_i \) is

\[
T_{r,i WP} = \frac{n_e(d - x_i \tan \theta)}{c} + n_o(d + x_i \tan \theta).
\]

(5)

Emerging from WP, orthogonally polarized reference and sample trains are able to interfere on the conditions that: (1) their initial group delays are compensated in WP to \( \Delta \tau \), and (2) their orthogonal polarization states are recombined to a single linear polarization state to yield interference fringes.

The latter interference condition is readily satisfied by means of the linear polarizer \( P_3 \). Then, by analogy of birefringent interferometers, the fringe pattern with the best visibility will be localized in the plane \( \Sigma \) shown by the dotted line inside WP in Fig.1 (lower right). This plane is the locus of snap points of the sample and reference rays that are deflected after WP apart from each other at the angle \( \varphi \).

The first of the preceding interference conditions implies that the best interference contrast is achieved in the vicinities of \( x_i \) and \( x_j \) on the plane \( \Sigma \) if

\[
T_{i WP} - T_{j WP} = T_{i WP} - T_{j WP},
\]

(6)

and

\[
T_{r,i WP} - T_{r,j WP} = T_{r,i WP} - T_{r,j WP}.
\]

(7)
Now, we define a theoretical interval of depth attainable in our scheme. After trivial algebraic manipulations of Eqs. (7), we have

\[ \tau_{M}^{i,j} - \tau_{M}^{i} = \frac{2(n_{i} - n_{j})(x_{i} + x_{j})\tan \theta}{c}. \]  

(8)

The left side of Eq. (8) represents the double time of traveling the depth interval \( L_{i,j} \) between interfaces \( i \) and \( j \). Introducing the mean refraction index \( \bar{n} \) of the sample medium within the \( L_{i,j} \), we easily obtain from Eq. (8)

\[ L_{i,j} = \frac{(n_{i} - n_{j})(x_{i} + x_{j})\tan \theta}{\bar{n}}. \]  

(9)

In the general case, any range \( a \leq x_{i} + x_{j} \) of the WP light aperture correlates to the following theoretical depth interval:

\[ L = \frac{(n_{i} - n_{j})a \tan \theta}{\bar{n}}. \]  

(10)

It should be emphasized more particularly that the true decoupling between arms is ensured by orthogonality of polarization states of beams transmitted through each arm. The beams reflected from various interfaces \( i, j \),... in the sample become linearly polarized after passing through the polarizing beam splitter PBS and then enter WP. Each of these partial beams travels the halves of WP either as an extraordinary or as an ordinary beam. These beams are not split by WP, but are deflected equally, and propagate as a whole beam. The latter cannot form by itself a spatially localized fringe pattern. In contrast to that, orthogonally polarized reference and sample beams cannot form by itself a spatially localized fringe pattern. In the WP, by some analogy with the Fresnel double-mirror experiment. Reflections between any virtual interfaces in the sample will contribute to this pattern, but incoherently. Therefore, the pattern will be devoid of autocorrelation artifacts that are intrinsic, e.g., to spectral domain OCT. This excludes additional corrections based on smart software or other contrivances.

Obviously, if the \( r \)-train together with the \( i \)- or \( j \)-train travel equal optical path lengths in MZI, they lead to a zero-order interference fringe in WPI. Location of this fringe corresponds to the equal wedge thicknesses in WP, i.e., positions where \( x_{i} = 0 \) or \( x_{j} = 0 \). In some practical cases, the zero-order interference fringe should be shifted from the center of the WP aperture. For that, it is necessary to change the relative thicknesses of the wedges or insert in front or behind WP, a birefringent plate, which is cheaper. At the design stage, a required position of \( L \) can be present beneath the sample surface by choosing the proper plate thickness. Also, the optical path length of the reference arm and the probing beam waist position should be optimized. Included in Eq. (10), values of \( \bar{n} \) for human skin and \( (n_{i} - n_{j}) \) for calcite crystals are well known within a wide spectral range.

Noteworthy, WP-type prisms based on three birefringent wedges have equivalent optical characteristics at small \( \theta \) and therefore provide better quality of the fringe pattern. A slight slope of the plane \( \Sigma \) relative to the faces of WP is less in the three- than the double- or the single-wedge devices. A considerable slope deteriorates a proper projection of the fringes onto the CCD plane. The slope can be eliminated in WP of special design, but we prefer to use standard componentry. The more practical solution is seen in slightly tilting a standard WP in the plane of beam splitting.

### 2.3 Time Division of Polarization Components

Still referring to Fig. 1, consider the more general case. Let the linearly polarized probing light after double-pass traveling the sample acquire both the horizontal \( I_{h} \) and the vertical \( I_{v} \) intensity components of the polarization state. Orientations of these components are denoted by \( (\|) \) and \( (\perp) \), respectively. In general, both sample birefringence and scattering influence the polarization state of the probing light. At submillimeter depths in normal human skin, linearly polarized incident light is maintained preferentially over circularly polarized light.

At larger depths, PS OCT with linearly polarized light may result in better detection sensitivity and reduction of systematic errors in birefringence measurement as compared with circularly polarized light. At submillimeter subcutaneous depths, especially with the use of PS OCT, contribution of scattering to polarization changes can be neglected. Regardless of preference for either polarization, all emerged PS OCT topologies are based on the well-known concept of two independent polarization channels. The channels are discriminated through a spatial separation or in combination with a software-based technique.

Instead of the spatial separation, we propose a temporal separation, i.e., time division of polarization components. The idea consists in a high-speed alternate transmitting of either orthogonal polarization component through the same optical channel to the same detector. For that, the bistable half-wave switch BHWS\(_{1}\) is introduced in the sample arm of MZI. As shown schematically in Fig. 2, BHWS\(_{1}\) based on the ferroelectric-liquid-crystal cell operates as an electrically switchable half-wave retarder. Diagrams of polarization states and optical axes and planes are represented in successive sections along the beam propagation from left to right. Reversing the polarity of dc low voltage applied to BHWS\(_{1}\), changes orientation of its principal axis exactly from one steady state at 0 deg to another state at 45 deg and back. Being sufficiently achronatic, BHWS\(_{1}\) transforms the incoming polarization state, elliptical in the general case, to an intermediate polarization ellipse. The latter holds the same shape but is the mirror of the incoming ellipse about the BHWS\(_{1}\) principal axis activated at the moment. BHWS\(_{1}\) in combination with PBS works both as a discriminator and an accurate discrete modulator of incoming polarization state components \( I_{h} \) and \( I_{v} \). The discrete modulation is more advantageous than a continuous one since it has less fringe washout during lateral scanning.

Functional testing of the commercial model LV2500-OEM (Displaytech, Inc.) pre-designed as BHWS\(_{1}\) gave us polarization modulation depth up to 27 dB for incoming light from SLD with the central wavelength \( \lambda = 827 \) nm of the spectral band \( \Delta \lambda = 42 \) nm. For this device, the typical time of transition between two steady optical states is specified as short as tens of microseconds, and the maximum permissible frequency of modulation is 20 kHz.
Thus, the mechanism of fringe formation for each of components $I_1$ and $I_\perp$ comes directly to the case we analyzed in the preceding section. It is also evident in Figs. 1 and 2 that to change by 45 deg the principal axis orientation of the additional switch BHWS$_2$, identical to BHWS$_1$ the phase of the reference beam will undergo the half-wave flip. Such modulation yields either in-phase or anti-phase fringes. At zero OPD, the in-phase fringes $F_{\text{in}}$ will exhibit positive peaks, and the anti-phase fringes $F_{\text{anti}}$ will exhibit negative peaks of a noisy interference signal $s$ superimposed on a dc background $B_{\text{dc}}$. The background $B_{\text{dc}}$ consists of a noncoherent stationary component and a constant bias caused, e.g., by dust particles. This background does not interfere and is independent of the polarization. In addition, it is assumed to be the same in both fringes $F_{\text{in}}$ and $F_{\text{anti}}$ at least during the period of modulation,

$$F_{\text{in}} = B_{\text{dc}} + s,$$

$$F_{\text{anti}} = B_{\text{dc}} - s.$$ 

(11)

(12)

As seen in Eqs. 11 and 12 when subtracting the anti-phase fringes $F_{\text{anti}}$ from the in-phase fringes $F_{\text{in}}$ the background $B_{\text{dc}}$ is suppressed and the interference signal $s$ is added. The resultant fringe pattern $F$ is expressed as the following:

$$F = F_{\text{in}} - F_{\text{anti}} = (B_{\text{dc}} + s) - (B_{\text{dc}} - s) = 2s.$$ 

(13)

Although this fringe pattern still contains the initial noise, suppression of the noncoherent stationary component and the constant bias promotes better fringe visibility and performance reliability. Seemingly, some analogy with a balanced detection can be drawn for such a technique.

3 Experimental Validation

3.1 Balancing of Arms

Noteworthily, some experimental findings are added to elaborate the initial design. As is known, the intensity of the beam backscattered from most biological tissues is a few orders of magnitude less than the incident beam intensity. That is why the reference arm intensity should be attenuated by two or more orders of magnitude to improve the signal-to-noise ratio. As seen in Fig. 1, the reference channel of MZI uses no traditional means to attenuate the reference beam intensity. This function is fulfilled successfully with the linear polarizer $P_3$. Basically, this optical element serves to combine by polarization the reference and sample beams that propagate in WP either as ordinary or extraordinary beams. On the other hand, smooth rotation azimuthally of $P_3$ allows reciprocal balancing of intensity between the arms with any ratio from 0 to 1. In particular, such an original technique enables easier control of the fringe pattern contrast.

3.2 Projection of Fringes

To increase still more the optical efficiency of our scheme, the projection system comprises cylindrical lenses $CL_1$ and $CL_2$ arranged as shown in Fig. 4. $CL_1$ projects on CCD the fringe pattern localized inside WP the fringe pattern “condenses” more light across the length of the linear CCD array. A spherical objective lens used instead of $CL_1$ would not give both an effective condensing and a sharp imaging with variable magnification. The reason is the intrinsic astigmatism of such a system. Since $CL_1$ and $CL_2$ are crossed at 90 deg, they form the so-called anamorphic lens system. It is the optimal solution in our particular case inasmuch as the fringe direction is orthogonal to the length of the linear CCD array. Figure 5 illustrates a fragment of the fringe pattern formed using an alignment red laser. The pattern is projected with fourfold magnification from the plane inside WP onto the plane of the expected location of the CCD. Magnification could vary optionally. For comparison, sizes of a single pixel are shown on the right side. The pattern looks somewhat distorted because it is taken using self-made magnifying optics and an amateur camera (Samsung KENOX S500). Nevertheless, the picture is evidence of the dramatically increased optical throughput due to making use of the anamorphic lens system. Also, such an approach may be helpful to the spectrometer design used in spectral domain OCT systems.

Fig. 2 Alternate transmitting the orthogonal polarization components of the incoming polarization state.

Fig. 3 “Condensed” fringe pattern (fragment).
3.3 “Rule of Thumb”

Figure 4 illustrates the hardware layout of the experimental setup based on the tandem interferometer assembled according to the optical schematic in Fig. 1. It is worthy of notice that the interferometer does not contain high-precision translational stages for lateral positioning, in contrast to the systems of spectral domain PS OCT with two optical channels. Nevertheless, due to exploiting a single optical channel and a single detector, the interferometer alignment was relatively straightforward, although additional efforts were required to manipulate bulk optics. On the other hand, bulk optics makes assembling of unique configurations in some respects more flexible.

The interferometer structure comprises commercially available optical components that are sufficiently achromatic and have high polarization contrast in the spectral range \( \Delta \lambda = 42 \text{ nm of SLD (Inphenix Corp.)} \). The central wavelength \( \lambda = 827 \text{ nm of SLD} \) is near to the maximal spectral response of the exploited linear CCD array, in particular, model AViiVA M4 (Atmel Corp.). The total number of pixels \( N_p = 4096 \), and the pixel width \( \delta p = 10 \mu \text{m with fill factor close to unity} \). As shown here, \( N_p \) is a crucial parameter that determines the theoretically attainable depth interval \( L \) given by Eq. (10). To establish a practical relationship between \( L \) and \( N_p \), a kind of rule of thumb can be derived from the following considerations. Using previous notations, the spatial period of fringes inside WP is expressed with the well-known formula:

\[
\delta f = \frac{\lambda}{\varphi},
\]

where \( \varphi = 2 (n_s - n_a) \tan \theta \) is the angle of beam splitting for WP. Combining Eqs. (14) and (10) yields

\[
L = \frac{\lambda a}{2 \pi \delta f}.
\]

Assume \( a = A \), where \( A \) is the full light aperture of WP projected with magnification \( M \) onto the full physical length \( D \) of the linear CCD array. Suppose also that \( \sigma \) is a spatial frequency of sampling defined as a number of pixels per fringe period \( \delta F \) of the projected pattern. Under these assumptions, the following obvious relationships are valid

\[
M = \frac{D}{A} = \frac{\delta F}{\delta f}, \quad N_p = \frac{D}{\delta p}, \quad \sigma = \frac{\delta F}{\delta p},
\]

and manipulation of Eqs. (15) and (16) gives finally

\[
L = \frac{\lambda N_p}{2 \sigma 
\]

Thus, the attainable depth interval \( L \) is determined only by the total number of pixels \( N_p \), the central wavelength \( \lambda \), the spatial frequency of fringe sampling \( \sigma \), and the mean refraction index \( \tilde{n} \) of the sample medium within this interval. The rule of thumb expressed with Eq. (17) is valid for estimation of the properly attainable value of \( L \), provided the WP aperture is effectively filled with the light beam and projected onto the full length of the linear CCD array.

3.4 Signal-to-Noise Considerations

Compare our solution with the interferometer introduced in L-OCT and based on the principle similar to Young’s two-pinhole experiment. Although our interferometer differs conceptually from the preceding, the common feature of both interferometers is that the fringe pattern is formed spatially and projected on a linear CCD array for further readout. An important point is that the attainable depth interval \( L \) obeys the same Eq. (17) for both interferometers. \( L \) is spatially evolved along the full physical length \( D \) of the linear CCD array. \( D \) is scanned electronically during the CCD readout time \( T \). It is equivalent to the mechanical scanning of \( L \) with some velocity \( V \). For L-OCT and time-domain OCT (TD, OCT), the approximate signal-to-noise ratios \( (SNR) \) in the case of shot-noise-limited detection of analog signals are expressed with similar equations:

\[
SNR_{L-OCT} = \frac{l_{coh} \Phi_s}{L T},
\]

\[
SNR_{TD-OCT} = \frac{l_{coh} \Phi_s}{V},
\]

where \( \Phi_s \) is the total photon flow incident on the CCD array from the sample arm. It is worthy of special notice that Eq. (18) is valid for our system as well. As is known, TD OCT may exhibit superior performance with respect to SNR, especially in turbid media. Therefore, in spite of the simplified consideration, the analogy with TD OCT still makes L-OCT and our solution promising for probing the nontransparent tissues. In general, PS OCT is able to achieve \( SNR \) sufficient for mapping cumulative (macroscopic) birefringence of skin with the reasonable resolution at submillimeter depth. \( SNR \) involvement of polarization-preserving fiber optics as well as the replacement, with time, of CCD by complementary metal-oxide semiconductor (CMOS) would contribute to our system performance.

3.5 Comparison of Theory and Experiment

Figure 5 shows the fringe pattern resulting from the interference between the recombined orthogonally polarized reference beam and sample beam in our experimental setup. The sample beam is reflected from the polarization insensitive sil-
ver mirror. While translating the mirror along the optical axis of the sample arm, the pattern shifted along the array. Making use of the anamorphic projection and the reciprocal balancing described earlier facilitated obtaining the distinct interference pattern superimposed on a dc background. The pattern is somewhat affected by air turbulences and parasitic interference fringes typical of a Mach-Zehnder interferometer based on bulk optics. Exploiting of single-mode fiber optics would suppress these artifacts and smooth the fringe pattern due to spatial filtering.

Replacing the silver mirror with a coverglass plate (Menzel Gläser GmbH) of thickness 148 µm, measured with a caliper resolution of 1 µm, yielded two separate fringe patterns spaced apart and produced by reflections from each of the plate surfaces. As expected, the patterns moved synchronously with shifting of the plate. The distance between the main peaks of the patterns remained the same. This is in accordance with the basic properties of the tandem interferometer revealed earlier. A standard two-element Wollaston prism (CVI Corp.) made of calcite was used as WP and had φ = 5 deg. The roughly estimated distance between the experimentally observable peaks gave the number of pixels \( N'_p \) = 2500 ± 20. The magnification \( M = 4.8 \) was calculated from the thick lens formula applied to the cylindrical lens \( CL_1 \) of model CKX038 (Newport Corp.). Using the evident relationship

\[
a = \frac{\delta p N'_p}{M},
\]

and substitution of Eqs. (14) and (20) into Eq. (15) yields

\[
L = \frac{\delta p N'_p \varphi}{2h M}.
\]

After substitution into Eq. (21) of the known numerical values and assuming \( n = 1.52 \) for the used coverglass plate, we obtained finally \( L = 149 \) µm. This is in a good agreement with the plate thickness 148 µm, which we gauged directly using the caliper.

From the fringe envelope subjected to bandpass filtering, we obtained \( I_{coh} \) of SLD. This was about 8 µm and correlates well with the theoretical value 7 µm evaluated for free-space from the spectral characteristics of SLD, i.e., \( \lambda = 827 \) nm and \( \Delta \lambda = 42 \) nm corresponding to full width at half maximum (FWHM) of the spectrum approximated with a Gaussian profile.

This decisive experiment is plain but indispensable evidence of the ability of the embodied interferometer to decouple between arms by polarization and under the given resolution to “sense at once,” without mechanical A-scans, any reflective interface fallen into the attainable depth interval. According to the derived rule of thumb, at \( \tilde{n} = 1.4 \) typical of the human skin and \( \sigma = 3 \), i.e., the one-and-half Nyquist frequency, the commercial cameras AViiVA with linear array formats 4k, 6k, and 8k are well suited for submillimeter depth intervals 0.4 mm, 0.6 mm, and 0.8 mm, respectively. The interferometer design allows presetting these intervals of interest further beneath the superficial layers, e.g., behind the epidermal-dermal junction of human skin.

To distinguish between the normal and depleted collagen contents in dermis only through measurements of \( I_1 \) and \( I_2 \), a phenomenological model of depolarization is evolved here.

### 4 Depolarization in Dermis

#### 4.1 Phenomenological Model

In dermal layers, collagen fibrils are packed in collagen bundles and have lamellae structure. The finely woven meshwork of reticular collagen fibers is typical of the upper dermis, whereas thick, coarse collagen bundles are the main feature of the deeper layers. In respect to crystal optics, collagen fibrils in skin are nothing but liquid-crystal structures with typical sizes varying from nano- to micrometric scales. Human skin biopsies show that birefringent crystalline fibers of collagen, however, are woven in dermis layers more or less chaotically but arranged mainly parallel to the skin surface. With relation to polarimetry and in particular PS OCT, a lamella can be considered as a linear tiny retarder in which slow and fast axes lie, respectively, parallel and perpendicular to the fibril direction. Noteworthy, known from the literature a birefringence model of PS OCT for human cornea is based on a stack of thin birefringent lamellae with two preferential, nearly orthogonal orientations, superimposed on a background of lamellae with random orientation.

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**Fig. 5** A snapshot of the spatial fringe pattern.
The lamellae in the human dermis have a much more disordered structure than in the cornea. Can a great many chaotically oriented collagen lamellae be considered as a single object that still displays birefringent properties? To answer this question, we referred to the theory of light depolarization in a statistical system of parallel crystal plates. The theory well interprets depolarization of light for crystallizing polyethylene of high density. In general, the lamellar microstructure of collagen in the dermis looks, in our thinking, rather close to crystalline morphology of such polyethylene although birefringence values differ almost by one order of magnitude.

Consider a single lamella of thickness $t$ and optical birefringence $\Delta n$, placed between crossed polarizer and analyzer. The optical axis of the lamella is oriented azimuthally at the angle $\alpha$ with respect to the polarizer. From the theory of compensators, provided absorption and spectral dispersion are negligible, the intensities of probing light with a center wavelength $\lambda$ transmitted through the analyzer parallel and perpendicular to the polarizer are in the ratio corrected for scattering,

$$\frac{I_{\parallel}}{I_{\perp}} = \sin^2 2\alpha R_i,$$  \hspace{1cm} (22)

$$R_i = \sin^2 \frac{\pi t \Delta n}{\lambda},$$  \hspace{1cm} (23)

where $R_i$ is the optical retardation of the individual lamella.

Figure 6 shows a statistical system of lamellae chaotically oriented azimuthally but arranged parallel to any cross section of the probing light beam. For such a system, the aforementioned theory yields the depolarization ratio $D$ of transmitted light

$$D = \frac{2I_{\parallel}}{I_{\parallel} + I_{\perp}} = 1 - \exp(-RN),$$  \hspace{1cm} (24)

where $R$ is a variable similar to that in Eq. (23) but averaged over the ensemble of $R_i$, and $N$ is the average number of lamellae encountered on the total path of the light beam with the cross section defined by the circle in Fig. 6. In the strict sense, the depolarization term $RN$ is not a unique function of material quantities involved in Eq. (23) but depends also on statistical distribution both of lamella sizes and concentrations.

For purposes of further analysis, still hypothesize the validity of decoupling the $RN$. Suppose that in Eq. (23) variables $t$ and $\Delta n$ are already averaged values, and substitute $R_i$ for $R$ in Eq. (24). Obviously, for measurement of $I_{\parallel}$ and $I_{\perp}$ with the PS OCT system shown in Fig. 1 the polarization plane of $P_1$ should be preset at 90 deg. A probing light beam returning from the dermis depth $z$ will undergo depolarization on the double rather than the single pass. Therefore, we have to intuitively suppose the validity also of the optical reciprocity principle. Further, assume that the collagen lamellae have average interspacing $s$ between layers. Suppose also that the average number of the lamellae per unit cross-section area of the probing beam is $c(z)$ and the beam radius is $r$. Now, there is no difficulty to come from Eqs. (23) and (24) to

$$D(z) = 1 - \exp \left[ -\frac{2\pi^2}{s} \sin^2 \left( \frac{\pi t \Delta n}{\lambda} \right) \int_{z_0}^{z} c(z)dz \right],$$  \hspace{1cm} (25)

where $z_0$ is the depth of the epidermis-dermis interface form which the dermis displays noticeable birefringent properties, i.e., has $c(z) \geq 0$. As shown here, the algorithm based on measurements $I_{\parallel}$ and $I_{\perp}$ is able to identify inferior margins of simulated tumor.

4.2 Estimation Algorithm

Consider a computational “phantom” simulating a dermis-like medium with realistic optical characteristics given in Table 1.

The values of parameters $s$ and $t$ are close to those given in Ref. 67 $z_0$, $c_o$, and $\Delta n$ are taken, Refs. 54, 68 and 69, and $\lambda$ and $r$ are typical of PS OCT. First, imitate a rise of lamella density with depth for normal dermis as

$$c(z) = c_o \left[ 1 - \exp \left( -\frac{z}{z'} \right) \right],$$  \hspace{1cm} (26)

In Eq. (24) $c_o$ is the maximal concentration attainable far beyond the depth of interest, and $z'$ is some depth in upper dermis with lamella concentration $c_o(1-1/e)$, where $e$ is the

Table 1: Design parameters of the computational dermis-like phantom.

<table>
<thead>
<tr>
<th>$\lambda$ (nm)</th>
<th>$r$ (µm)</th>
<th>$s$ (µm)</th>
<th>$t$ (µm)</th>
<th>$z'$ (µm)</th>
<th>$z_0$ (µm)</th>
<th>$c_o$ (mm$^{-2}$)</th>
<th>$\Delta n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>827</td>
<td>7.5</td>
<td>16</td>
<td>6</td>
<td>200</td>
<td>100</td>
<td>$3 \times 10^4$</td>
<td>$3 \times 10^{-3}$</td>
</tr>
</tbody>
</table>
base of the natural logarithm. The depolarization ratio $D(z)$ calculated from Eq. (25) for this case is shown by the upper curve in Fig. 7.

Assume that a simulated tumor invades in dermis to the depth of 500 μm and depletes the lamella of the level of $e(z)/e$ homogeneously throughout the tumor volume. The homogeneity is quite an admissible assumption because, e.g., the early melanoma has a compact tumor and on the other hand, a tumor of smaller size has more homogeneous structured. The choice of the $1/e$ levels is caused just by simplicity of the exponential presentation. For this “depletion” case, the depolarization ratio $D(z)$ is reduced considerably, as shown by the bottom curve in Fig. 7. In addition the curve of $D(z)$ undergoes an upward bend at the inferior margin 500 μm of the simulated tumor. Qualitatively, these peculiarities resemble the increase in cumulative birefringence and the similar bend that were demonstrated with PS OCT by Strasswimmer et al. for basal cell carcinoma of nodular type and the similar bend that were demonstrated with PS OCT by Tougbaev et al. for basal cell carcinoma of nodular type.

In the optical schematic shown in Fig. 1, the sections of optical paths defined by {P1-NPBS, {BHWS1-PBS}, and {NPBS-M1+4-BHWS2} are devised specifically with the possibility of exploiting polarization-preserving fibers in minimal combination with bulk optics. These fibers will be integrated in a flexible, advantageously shielded cable. The cable should couple a handheld head of the optical probe to the main unit of the system. However, this is a challenge to our further research and development. One of the expected engineering issues is the coupling efficiency of interconnections between small-core fibers and bulk optics. Fortunately, commercially available polarization-maintaining fibers possess sufficient stability of polarization-preserving properties with respect to bending, twisting, tension, and temperature changes, which was proved in the severe field environment. In addition to these properties, single polarization fibers combine high polarization extinction with an exclusively wide spectral band. This makes them prospective candidates for fiber-based PS OCT.

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References


