Towards simultaneous Talbot bands based optical coherence tomography and scanning laser ophthalmoscopy imaging

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Abstract: We report a Talbot bands-based optical coherence tomography (OCT) system capable of producing longitudinal B-scan OCT images and en-face scanning laser ophthalmoscopy (SLO) images of the human retina in-vivo. The OCT channel employs a broadband optical source and a spectrometer. A gap is created between the sample and reference beams while on their way towards the spectrometer’s dispersive element to create Talbot bands. The spatial separation of the two beams facilitates collection by an SLO channel of optical power originating exclusively from the retina, deprived from any contribution from the reference beam. Three different modes of operation are presented, constrained by the minimum integration time of the camera used in the spectrometer and by the galvo-scanners’ scanning rate: (i) a simultaneous acquisition mode over the two channels, useful for small size imaging, that conserves the pixel-to-pixel correspondence between them; (ii) a hybrid sequential mode, where the system switches itself between the two regimes and (iii) a sequential “on-demand” mode, where the system can be used in either OCT or SLO regimes for as long as required. The two sequential modes present varying degrees of trade-off between pixel-to-pixel correspondence and independent full control of parameters within each channel. Images of the optic nerve and fovea regions obtained in the simultaneous (i) and in the hybrid sequential mode (ii) are presented.

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1. Introduction

Due to a higher sensitivity and speed than their time domain (TD) counterpart [1,2], spectral domain (SD) methods dominate the OCT technology of eye imaging [3,4]. The SD-OCT methods produce fast A-scans, which are used to create real time cross-section (B-scan) images.

For several reasons (detailed below), an en-face image is also required when imaging the retina in the human eye. Besides the guidance of OCT examination, such an image can also be used to correct eye motions in the OCT data [5]. When the fundus image presents sufficient resolution, it may serve clinical assessment in conjunction with the B-scan OCT image. For instance, a commercial system from Topcon [6], uses an SLR digital camera to provide a full field image of the fundus. As another possibility, based on a summed voxel projection (SVP) [7], the strength of interference of all pixels along the depth within each A-scan needs to be summed up to produce a brightness value. Such a procedure is the most common method to produce a fundus-like image using SD-OCT technology, this image then being presented static to the user as shown in [8].

Another available option is to employ scanning laser ophthalmoscopy (SLO), and integrate such a system within a separate OCT set-up. This requires diverting some light from the returned beam from the retina, in a way similar to that previously practiced by the time domain technology of OCT combined with SLO [9–11]. Optos’ OCT/SLO instrument [12] uses the same flying spot principle used in OCT imaging to sequentially generate an SLO image, effectively having both channels sharing the same optical scanning head [13]. A different principle was used in [14], where separate transversal scanners are used for a swept source OCT channel and for a SLO channel operating at different wavelengths via a dichroic splitter. The associated SLO channel provides imaging as well as it can be used as the tracker itself. A similar concept was implemented in a hand held probe, appropriate for imaging subjects with less stable fixation, such as children [15]. A SLO channel using a line camera and a galvo-scanner was added to a B-scan time domain OCT to create a sequential SLO/OCT system [16]. Sequential production of B-scans and confocal microscopy images has also been reported by means of fluorescence-based microscopy [17, 18]. By interlacing a spectrally-encoded confocal SLO frame and an OCT B-scan in alternate fashion [19], a fundus image is generated during the same time interval required to grab a B-scan.

All the methods and systems mentioned above allow the user to relate the features seen in the en-face image with the features in the OCT B-scan, with various degrees of pixel-to-pixel correspondence. These approaches show the interest for presenting an en-face image together with the SD-OCT investigation delivering B-scans.

In this paper, we refer to OCT using a broadband source and a spectrometer, from now on referred to as Sp-OCT technology [3,20], inspired from the technology of spectral interferometry used for sensing [21]. An improved solution for the display of dual images [Sp-OCT B-scan]/[C-scan SLO] is presented, based on a Talbot bands (TB)-configuration [22]. In order to implement a conventional Sp-OCT configuration, a single splitter is needed to split light into a sample and reference path. However, this has the disadvantage of sending light back from the reference path towards the broadband source, which may lead to noise or even destroy the
optical source. Therefore, an isolator or a circulator may be needed to protect the source [20]. As another possibility, a two-splitter configuration may be used, where the reference path is recirculated, similar to that employed in [9]. In such a configuration, to add a SLO channel a 3rd splitter would be necessary to divert some of the sample light towards a SLO receiver. A better solution is presented in this paper, where the SLO channel picks up its signal from an otherwise wasted beam of light when the two interferometer arms are reunited by the second splitter. The configuration proposed has the advantage of a more efficient use of power originating from the eye, when compared with configurations having recirculation of the reference path, which would require a 3rd splitter to tap signal from the sample.

A second advantage of the solution presented is that of the improvement in the OCT channel sensitivity profile versus depth. A TB configuration allows a fine control of the position and span of the OCT channel’s sensitivity profile over the optical path difference (OPD) axis [23].

In conventional implementations of Sp-OCT technology [3, 20], the spectrometer performs spectral analysis of the interference product delivered by the interferometer. In such set-ups, the order of the two operations, interference and diffraction (dispersion) [22] is interference first. The two beams returning from the interferometer’s arms travel as a single beam along the same path between the interferometer and spectrometer. In a TB configuration, however, the sample and reference beams from the interferometer travel along distinct paths. A lateral offset is introduced between the two beams in their way towards the spectrometer, so that the projections of the two beams on the diffraction grating (or prism) are not fully overlapped. The non-overlapping parts of the beam footprints are diffracted before they are interfered. This reversal of order of operation between interference and diffraction determines the characteristics of TBs [22]. One of such characteristics is a shift of the OCT visibility profile V (OPD) away from a symmetric curve around OPD = 0 [24], the profile being determined by a factor CTB(OPD). This is due to the tilt of the wavefront after diffraction, which alters the overlap of the wave-train lengths after diffraction, as explained in [22]. A rigorous description of the CTB term is presented in [25] as the correlation of the spatial power distribution within the two beams. If the two footprints are not fully overlapped, as stated earlier, the CTB profile will no longer be symmetric in relation to the OPD = 0 axis and therefore either the positive or the negative OPD branch will be more attenuated than the other one, as shown in [23, 26]. It was shown [27, 28] that the visibility profile V (OPD) is described by:

\[ V(\text{OPD}) = C_{TB}(\text{OPD}) \left( \frac{\sin \frac{\xi}{\xi}}{\xi} \right)^2, \]

where \( \xi = \frac{\pi}{2} \frac{\text{OPD}}{Z_{\text{max}}} \) denotes the depth normalized to the maximum imaging range \( Z_{\text{max}} \), [28, 29], given by:

\[ Z_{\text{max}} = \frac{M \lambda_0^2}{4 \Delta \lambda}. \]

In (2), \( \Delta \lambda / M \) is the line camera pitch (\( M \) is the number of pixels used to photodetect the spectrum with bandwidth \( \Delta \lambda \)), and \( \lambda_0 \) is the central operating wavelength. The amount of sensitivity increase at a depth different from zero due to the shift of the \( C_{TB} \) profile depends on the relative width of the \( \text{sinc} \) profile \( \left( \frac{\sin \frac{\xi}{\xi}}{\xi} \right)^2 \), which in turn depends on the resolution of the spectrometer employed for the detection of the channeled spectrum. In most conventional Sp-OCT configurations, a fiber-based directional coupler is used to combine the two beams and produce their interference. In a Talbot bands configuration implementation, a bulk beamsplitter is employed to route the two beams towards the spectrometer, as the two beams need to
be spatially separated. It is this approach which then allows us to retrieve the confocal (sample) signal separately.

2. Experimental set-up

The dual channel Sp-OCT/SLO set-up is depicted in Fig. 1. Light from a super-luminescent diode $SLD$ (Superlum SLD-381-HP1-DIL-SM-PD, Cork, Ireland - central wavelength $\lambda_0 = 830$ nm, and spectral bandwidth $\Delta \lambda \approx 20$ nm) is directed towards the two interferometer arms (reference and sample) via a fiber-based 80/20 directional coupler $DC$ (AC Photonics, Santa Clara, CA, US). Given the SLD’s spectral bandwidth and central wavelength, an optical axial resolution in depth measured in air of approximately 15 $\mu$m results.

The 20% fraction of the initial power directed to the sample arm traverses a galvo-scanning head $SXY$ (Criel Instruments Galvoline G1432, Italy), comprising a line scanner $X$ and a frame scanner $Y$, where the former deflects the beam horizontally and the latter vertically. The resulting beam scans the retina angularly, via lenses $L_2$ and $L_3$ of focal lengths $f[L_2] = 7.5$ cm and $f[L_3] = 3$ cm being employed to reduce the beam diameter at the pupil eye to $\approx 3$ mm.

Light backscattered by the retina and light directed through the reference arm are re-united at the bulk beam-splitter $BS$, which features an 80/20 splitting ratio. 80% of the power returning
from the sample arm is directed towards the spectrometer in the OCT channel and the remaining 20% towards the SLO channel (forming the SLO en-face image). Similarly, the reference arm also sends 80% of its power through the beamsplitter. The gap between the two beams (adjustable by shifting the reference arm launcher C5 using TS1) required by TB implementation secures sufficient spatial separation of the sample and reference beams to enable the multimode fiber MMF to select mainly the sample beam through the collimator C6. For better attenuation of the stray signal caused by the edge of the strong reference beam, a specially-devised spatial filter is implemented. This consists of a pair of opaque screens introduced before and after the beam-splitter along the path of the reference beam. These screens are attached to translation stages (TS2/TS3) so that their position can be adjusted with micrometric precision. The opaque screen attached to TS2 trims the edge of the reference beam on the side of the sample beam directed towards C6. Due to diffraction registered at the edge of the beam caused by this screen, some light is directed towards C6, therefore a second screen attached to TS3 is necessary. For better rejection of the reference signal, this is pushed towards the center of the collimator C6, and a small fraction from the edge of the sample beam is blocked, as shown exaggerated in the inset in Fig. 1. This introduces a small attenuation on the SLO channel, quantified by measuring the SLO signal with and without TS2, of 2.3 dB. By screening the reference arm beam with TS2 by \( \sim 1 \) mm, a 3 dB attenuation of the reference power in the OCT channel is introduced. This figure was measured by evaluating the signal due to the reference beam with and without TS2. When using a mirror as a sample, the ratio between the sample signal power and leaked reference power in the multimode fiber leading to the SLO channel was of 35 dB. Without the two screens in place, this ratio reduces to 5 dB.

Due to the screen attached to TS1, the distribution of power in the transversal section of the reference beam results in a trimmed Gaussian curve. This affects the symmetry of the OCT sensitivity profile versus OPD, as documented in [24, 27]. However, the deviation of the profile from Gaussian has little influence on the final visibility profile due to the fact that the trimmed portion coincides with a wing of the sample beam (which carries less power) and not with the central part of the sample beam.

Spectral analysis of the OCT signal employs a diffraction grating (whose grooves are orthogonal to the plane of Fig. 1) working in transmission TG (Wasatch Photonics, Logan, UT, US), with 1200 lines/mm blazed at 830 nm, a CMOS line camera (Basler sprint spl-4096km, Ahrensburg, Germany) and an achromatic lens L1 to focus the diffracted light onto the camera array. The diffracted beam covers \( \sim 1 \) cm over the array, which corresponds to 1024 out of the available 4096 pixels. This reduced number of pixels was employed to enable the camera to operate at high line rates (up to 100 kHz).

For each lateral scan, 1024 spectral points with 12-bit levels are buffered, followed by standard software-based processing (re-sampling in \( k \)-domain, zero padding and inverse FFT). After the FFT is performed on the input data, the number of pixels in depth in the B-scan OCT image is half of this number. For \( N_t \) pixels along the line (\( O_x \)), the image has a size \( N_t \times 512 \) pixel (width x depth).

The SLO channel is equipped with an avalanche photo-diode (APD, Hamamatsu C5460-01, Japan), with a cut-off frequency \( f_c = 100 \) kHz. Its output is then digitized by an analogue-to-digital converter within a DAQ (National Instruments PCI-6110, Austin, TX, US). The reflected fraction of the sample arm power (at the beamsplitter BS) is directed to the APD via collimator C6 and multi-mode fiber, MMF.

The SLO channel delivers an en-face (constant depth scan or C-scan) comprised of multiple T-scans (lateral reflectivity profiles). Each of these T-scans is acquired during the active half-periods of the line scanner (\( x \) axis). The OCT channel delivers a B-scan (cross section) OCT image, which in turn is comprised of several A-scans (axial reflectivity profiles) taken during
Fig. 2. Schematic diagram of the control system of the OCT/SLO set-up to achieve the three modes of operation. The multi-function data acquisition card (DAQ) and the External Function Generator deliver signals to the X-scanner and the Y-scanner via the Scanner driver box. The Switch box is used in the sequential modes only. The detected spectra from the CMOS camera are sent through a Camera Link bus to the image acquisition (IMAQ) card while the analog SLO signal is sent to the DAQ card.

The control system of the OCT/SLO set-up is depicted schematically in Fig. 2. The workstation PC is equipped with two distinct cards. The DAQ drives the two transversal scanners and acquires the SLO signal to produce the SLO image. The IMAQ card (National Instruments PCIe-1429, Austin, TX, US) interfaces the CMOS line camera in the spectrometer with the PC, thus acquiring the spectra which will yield the OCT image.

The line scanner $X$ is driven by a triangular or saw-tooth waveform, $x_{sc}$, generated by the control software. The frame scanner $Y$ is driven by a saw-tooth signal (90% duty cycle), produced by an external function generator (Hewlett-Packard 8116A, Palo Alto, CA, US) and triggered by a computer via one of the DAQ’s digital ports (TTL$_{y}$). The external function generator is employed to synchronize the LabVIEW frame acquisition loop with the saw-tooth waveform generation.

When a single $Oxz$ frame is needed (B-scan), the $y$ scanner is held stationary and a DC offset is applied to it, controlled by the DAQ output $y_{sc}[DC]$. This allows the user to map the position of the cursor placed over the en-face image to the corresponding B-scan being acquired, depending on the operating mode chosen, as it will be presented over the next section. Switching between the two signals, saw-tooth for the SLO regime and DC in the OCT regime is performed via a Switch box, whose output is selected via the TTL$_{y}$ signal. This signal is generated in software and channelled via the DAQ’s digital port (sw) towards the switch box.

Furthermore, the DAQ also drives the buffering of the spectra which will be used, after their FFTs, to form the B-scan, where the succession of A-scan acquisitions is controlled via the TTL signal (TTL$_{x}$).

3. Timing and acquisition speed constraints

The operation of the OCT/SLO instrument depends on a set of constraints introduced by the hardware available at present, whose parameters can be manipulated to trade-off resolution by speed and vice-versa.

With regards to the OCT channel detection, the CMOS camera allows various acquisition settings [20]. The absolute maximum line rate attainable by the CMOS camera used is 312 kHz, however such figure is only possible when several of the signal-enhancing features of the
camera are turned off, such as vertical binning – which enables the two 4096 × 10 μm CMOS lines to effectively behave as a single 4096 × 20 μm line – and also when reading a smaller subset of the whole 4096 pixel array [20]. With such line rates, the signal-to-noise ratio (SNR) will diminish, hence the choice of the line rate will necessarily introduce a trade-off between speed and sensitivity. Throughout all the experimental work carried out, the line rate was chosen to be in the range of 50-100 kHz, which enabled several of the aforementioned features of the CMOS camera. This determines an acquisition time per spectra of \( \delta t_{OCT} \approx 10 - 20 \mu s \).

The inertia of the galvo-scanners SXY constrains the speed of lateral scanning as well. Therefore, the line scanner’s triangular waveform is limited to 500 Hz to prevent heating and reliability issues [30].

Lastly, there is the issue of the lateral resolution. Let us say that the Airy disc diameter of the beam focused on the retina is \( D_0 \). Then the lateral image size is \( \Delta X = D_0 N_x \); \( D_0 \) can be approximated by \( 1.22 \frac{f \lambda_0}{D} \), where \( f \) is the focal length of the eye, \( D \) is the scanning beam diameter and \( \lambda_0 \) is the central operating wavelength. For an eye length of \( f \approx 25 \) mm, using \( \lambda_0 = 840 \) nm and \( D = 3 \) mm, \( D_0 \approx 8.5 \) μm. Considering the aberrations of the eye, \( D_0 \) can be approximated as \( \approx 10 \) μm. Given the line scanner’s frequency of 500 Hz, this means that a half-period will take 1 ms. Assuming a CMOS camera line rate of 100 kHz, this means that only \( N_S = 100 \) adjacent A-scans can be retrieved during one half-period, which effectively limits the lateral range to \( N_S D_0 = 1 \) mm. Any scanner amplitude setting which would project a raster scan with a larger span than that will under-sample the object in terms of the optical resolution. In such a case, there will be more than a single Airy disc diameter within each electronic pixel, i.e. within each A-scan.

The SLO channel also introduces a limit in the lateral size, albeit larger than that imposed by the OCT channel: the APD used has a bandwidth \( f_{3dB} \) of 100 kHz, which determines a rise time \( t_{rise} = 3.5 \mu s \) for the impulses at the APD output. For the same line scanner speed (500 Hz) as above, this allows \( N_S[SLO] = 280 \), i.e. an increase in the SLO lateral size. This may be of interested if higher resolution SLO images are desired, even without pixel-to-pixel correspondence.

Taking all these constraints into account we have devised three modes of operation, with varying degree of pixel-to-pixel correspondence and allowable acquisition times.

### 3.1. Simultaneous, small lateral size

In this mode, the system acquires an OCT B-scan frame during each SLO frame acquisition (Fig. 3(i)). The B-scan OCT image has pixel-to-pixel correspondence with a selected line placed over the SLO image by a cursor (whose \( y \) coordinate is used to select the instant when the B-scan is buffered amongst all the SLO T-scans), and the frames in the two channels refresh simultaneously. Pixel-to-pixel correspondence (Fig. 3(iii)) means that the two channels have the same lateral size, that is \( N_x[SLO] = N_x[OCT] = N_S \). Moreover, since a square aspect ratio was chosen, \( N_x = N_y \), taking into account that \( N_x = 100 \) for \( \delta t_{OCT} = 10 \mu s \) it means that the two frames (SLO: 100 × 100 pixels; OCT: 100 × 512 pixels) can be refreshed in \( T_e = \frac{1}{2} \cdot T_s N_S = 100 \) ms, i.e. at 10 Hz (the effective frame rate is closer to 8 Hz due to the signal processing time). The \( \frac{1}{2} \) factor in the expression for \( T_e \) stems from the fact that both ramps of the triangular scanning waveform are employed in the imaging process, which means that during the complete period of the \( x \)-scanning waveform (\( T_e = 2 \) ms) two OCT/SLO frames are acquired.

This choice of image size enables simultaneous imaging with pixel-to-pixel correspondence in the two channels at high refresh rates. Due to the imposed short lateral size, this mode of operation is suitable for small size imaging of the eye, imaging photoreceptor cells, as presented...
Fig. 3. Schematic description of the various modes of operation implemented. (i) and (ii) Sequence of frames in the three modes of operation, where the green shadows show the frame refresh period, and the orange glow shows the instants when the system switches between the two regimes, if applicable. (i) Simultaneous mode of acquisition: the two frames, OCT and SLO are acquired and refreshed at the same time: illustration of different vertical positions in the SLO image where the OCT B-scan is selected from by varying $Y_{\text{DC}}$: a single OCT B-scan is captured, even though more can be buffered if necessary; (ii) Hybrid sequential and sequential “on-demand”: the system is toggled between the two regimes (SLO and OCT), and signal is acquired in each regime on separate time intervals; the toggle is automatic in the hybrid sequential mode or performed manually in the sequential “on-demand” mode. In the hybrid mode the two images are refreshed at the same time, even though they are not acquired simultaneously. (iii) and (iv): Scanner waveforms ($x$ and $y$) and illustration of integration time on pixels within the spectral acquisition events, $\delta t_{\text{OCT}}$, (orange rectangles) each leading to an A-scan and integration time on pixels within a T-scan, $\delta t_{\text{SLO}}$ (brown rectangles) in all three modes. (iii) Simultaneous mode of acquisition; (iv) Hybrid sequential and sequential “on-demand” modes of acquisition.
in [31], with or without adaptive optics. Furthermore, if lateral resolution is not a major concern, the image size can be increased and such a mode can also be used as an assistive technique for retinal tracking, where only the major features are required (e.g. fovea, optic nerve) in order to supply information to the tracking algorithm, or to suppress motion artifacts already present in the images. The increased frame rate (up to \( \approx 8 \) Hz) makes this mode of operation tolerant to movement.

However, if a larger lateral size (\( N_x > N_0 \)) is desired, or if the SNR needs to be increased, this mode is no longer suitable. Achieving the same lateral size in both channels is only made possible by accepting signals of different time duration. This leads to two additional possible modes of operation, where the system operates in a single regime at any given time (Fig. 3(ii)).

### 3.2. Sequential “on-demand”

In this mode of operation (Fig. 3(ii)), the system continuously refreshes the SLO frames until the user switches the system to the OCT B-scan regime, at which moment the last SLO frame is frozen for guidance of the OCT imaging. During the SLO regime (lasting \( T_y[SLO] \)), the frame scanner is driven by a ramp which is part of a saw-tooth waveform, generated by \( FG \), with a 90% duty cycle to minimize the return dead time (Fig. 3(iv)). When the system is switched to the OCT regime, the frame scanner is stopped at a vertical position selected by \( Y_{sc}[DC] \), determining the \( y \)-coordinate of each B-scan being displayed. Furthermore, the speed of the line scanner is slowed down to allow for a larger \( N_x \) and for an increased \( \delta t_{OCT} \). Considering the maximum number of pixels \( N_x = 280 \) allowed by the APD bandwidth, the line scanner speed in the OCT regime is reduced to \( f_x' = \frac{N_y}{N_x} \cdot f_x = \frac{100}{280} \cdot 500 = 180 \text{ Hz} \). For longer spectrometer exposure times, even lower frequency \( f_x' \) values are needed.

This mode of operation presents the obvious drawback of less tolerance to eye movements, as the SLO image is no longer refreshed. However, this mode of operation allows for more freedom in setting the lateral image size and the OCT channel parameters, since the timing of each frame is not tied to the time constraints of the SLO channel. In essence, the system operates in the two regimes at independent frame rates, which may make it flexible to perform an array of clinical scenarios, where image quality and size in both channels are more important than the exact correspondence between the two images.

In terms of pixel-to-pixel correspondence, some variations result from a regime to the next due to the change of image size when switching from 500 Hz to 180 Hz ramps applied to the line scanner. This can be compensated by a corresponding decrease in the voltage \( V_x \) applied to the line scanner in the OCT regime in comparison to the value applied in the SLO regime, correction that can be accurately worked out using the position sensing signal delivered by the scanning driver board.

### 3.3. Hybrid sequential

In this mode the pixel-to-pixel correspondence between the OCT and the SLO frames is improved by automatically switching repetitively between the two modes, as shown in Fig. 3(ii).

For better SNR in the OCT channel, the CMOS camera’s integration time was increased to \( \delta t_{OCT} = 20 \mu s \). Furthermore, an averaging feature of OCT frames is also incorporated, which will determine an adjustable time for the system in the OCT regime, \( T_x[OCT] \) depending on the number of OCT images \( \Lambda \) to be averaged (Fig. 3(iv)). The time in the OCT regime increases linearly from \( T_y[OCT] \) (for a single OCT acquisition) to \( \Lambda \cdot T_y[OCT] \) when \( \Lambda \) frames are acquired to be averaged, thus yielding a better SNR. The averaging process in the OCT regime may not affect the toggle time if \( \Lambda \cdot T_x[OCT] \) is kept lower than \( T_x[SLO] / 10 \). For \( N_y = 100, T_y[SLO] = 200 \text{ ms} \), which means that \( 1 < \Lambda < 10 \). For \( N_y = 280, T_y[SLO] = 560 \text{ ms} \), so \( 1 < \Lambda < 28 \).
Fig. 4. Relation between the number of pixels, $N_x$, determining the lateral image size and the mode of operation applicable. The red shaded region corresponds to the settings which allow pixel-to-pixel correspondence between the OCT and SLO images, which is limited at $N_x = N_S$. Lateral image size is calculated using $N_x \cdot D_0$, with $D_0 \approx 10 \mu m$.

For larger SLO images, $A$ could be even larger without affecting the toggle time.

3.4. Lateral size constraints

Depending on the mode of operation chosen, the maximum lateral image size attainable without loss of resolution varies according to the graph in Fig. 4. The line across the plot corresponds to the special case where the lateral pixel size matches the Airy disc diameter $D_0$. Above the line, the electronic pixel size is smaller than the Airy disc diameter, i.e. the system over-samples both OCT and SLO signals. Below the line, the Airy disc diameter $D_0$ is larger than the electronic pixel size, i.e. the system under-samples the signals. Note that the line scanner period is maintained at 1 ms for all SLO operations, but it might be modified during the OCT regime, depending on the mode of operation chosen.

True pixel-to-pixel correspondence (red shaded region) is only attainable when operating in the simultaneous mode, which limits the lateral image size to less than 1 mm.

A degree of pixel-to-pixel correspondence is still achievable for $N_S < N_x < N_{SLO}$, i.e. for $100 < N_x < 280$, if correction of lateral image size is made to compensate for the swing variation of the $x$-scanner with the frequency of the applied signal.

Over $N_x = 280$, the APD starts behaving as a low-pass filter due to its finite rise time, hence the line rate has to be reduced in order to maintain the lateral resolution whilst allowing for a larger lateral size. This will necessarily have an impact on the frame refresh rate, making the system more prone to motion-induced artifacts.

4. Demonstration of the working principle

Figure 5 (i) shows the sensitivity profile versus OPD. The launcher $TS1$ was laterally moved by $\sim 0.25$ mm to create the gap necessary for TB implementation, which determines a shift of the peak of sensitivity from $OPD = 0$ to $-1.8$ mm as shown by the red circles and green triangles. The TB sensitivity profile conserves its width from the non-Talbot band case, but the maximum sensitivity reduces by about 2 dB. However, at larger depths, the gain exceeds 6 dB. The green curve shows the TB sensitivity versus OPD with the screen on $TS2$ in place. As demonstrated in [25], the sensitivity profile is given by the correlation of the power distribution within the footprints of the two beams incident on the diffraction grating. Due to diffraction on the screen edge, the power distribution of the reference beam is changed to a wider footprint.
with secondary lobes. This was documented in [27], where a screen was also used to modify the distribution of power in the reference beam across the grating. Therefore, a slight improvement of the sensitivity results.

The sensitivity was measured following the procedure described in [32] for the non-TB case (black curve, Fig. 5) and OPD set at $-1\text{ mm}$. For the two integration time values used in this study, $\delta t_{\text{OCT}} = 10$ and $20\ \mu\text{s}$, sensitivities of $82\ \text{dB}$ and $87\ \text{dB}$ were obtained, respectively.

Following the system characterization, several in-vivo retinal images were acquired from the eye of one of the authors (AP), covering different features in the eye: the foveal region and the optic nerve region. Ethical approval was obtained from the Faculty of Sciences’ Ethics Committee. Power to the eye was less than $750\ \mu\text{W}$, in accordance with standards [33].

### 4.1. Dual SLO/OCT retinal images

Figure 6 presents images obtained with the system running under the simultaneous mode refreshing at $3\ \text{Hz}$. The line scanner is driven with a triangular signal of period $2\ \text{ms}$ and the camera integration time is $10\ \mu\text{s}$. Two possibilities are presented, small size imaging with a lateral pixel size less or equal to the optical transversal resolution, constrained by the limited number of lateral pixels achievable as explained above, and large size imaging, where the system still operates with the same number of pixels and so the images are under-sampled.

In each box in Fig. 6, SLO frames are shown at the top with a resolution of $100 \times 280$ pixels (width $\times$ height). OCT frames are shown at the bottom row with a resolution of $100 \times 150$ pixels (width $\times$ depth, cropped from $512$ axial pixels to emphasize the region under analysis).

In Fig. 6(i) to 6(iii) the SLO images are about $500 \times 500\ \mu\text{m}^2$, therefore the pixel mesh is denser than the actual optical resolution of the system. In (i) the edge of the optic nerve head is shown.

In Fig. 6(ii), the volunteer looked halfway between the fovea and the optic nerve. The small size imaging allows distinguishing individual photo-receptors when the eccentricity of the location on the retinal image exceeds $5^\circ$. For such eccentricity, the cone spacing is larger than $\approx 10\ \mu\text{m}$ [34]. In (iii) photo-receptor cells are still visible, along with the choroid layer (yellow arrow).
Fig. 6. Retinal images obtained while running the system in the simultaneous mode of operation at a frame rate of \( \approx 3 \) Hz. SLO frames (top image in each frame) are 100 × 280 pixels and OCT frames (bottom image in each frame) are 100 × 512 pixels (here cropped to 100 × 150 pixels to emphasize the region under analysis). (i) edge of the optic nerve head, lateral size \( \approx 500 \times 500 \) μm\(^2\); (ii) region between the optic nerve and the fovea, in an area featuring larger photo-receptors (\( \approx 10 \) μm), lateral size \( \approx 500 \times 500 \) μm\(^2\); (iii) pair of SLO and OCT images (lateral size \( \approx 500 \times 500 \) μm\(^2\)) featuring a blood vessel; the chorioid (yellow arrow) is visible below the nerve fiber layer; (iv) optically under-sampled OCT image of the area between the foveal region and the optic nerve, lateral image size \( \approx 2 \times 2.5 \) mm\(^2\); (v) optically under-sampled OCT image of the optic nerve, the region in focus is the shallower retinal layer, lateral size \( \approx 1.5 \times 0.8 \) mm\(^2\). The OCT B-scans correspond to the location of the horizontal lines overlaid on the SLO C-scans.
Larger values for the lateral image size were also considered. In Fig. 6(iv), the line scanner was driven with $\approx 600 \text{ mVpp}$ determining about 2 mm lateral size. This size is larger than that obtained by multiplying the assumed Airy disc diameter of $\approx 10 \mu\text{m}$ with the number of transversal pixels $N_x = 100$, so the image is obviously under-sampled in the lateral direction, as commented above in connection to Fig. 4.

Figure 6(v) contains an optically under-sampled OCT image as well, since the lateral size is over 1 mm. The focal region was on the shallower retinal tissue, hence the corresponding OCT profile only maps the $x$-coordinates where the optic nerve is situated.

Features are sufficiently well seen in both columns, however due to the high speed of the camera, the OCT images are noisy. Even so, contours and main layers are easily identified at this frame rate.

Figure 7 features the results obtained with the OCT/SLO set-up operating in hybrid sequential mode. During the OCT regime, the lateral scanning duration of the line scanner is 20 ms and the camera integration time was increased to 20 $\mu\text{s}$, which enabled us to use more pixels in the lateral dimension of the OCT B-scans. Images of the foveal region and of the optic nerve are presented. The lateral image sizes considered ranged from 2.6 to 5 mm. Again, two image size values are employed: a medium lateral size of 2.6 mm, where the lateral pixel size in the SLO image is of the same order as the optical transversal resolution and a larger lateral size of 5 mm, where the SLO images are under-sampled. For both sizes, the same number of lateral electronic pixels $N_x \approx 500$ is used in both OCT and SLO regimes.

Figures 7(i) and 7(ii) differ mostly on the location of the focal point (in depth) - the former features a better defined retinal surface, whereas the latter emphasizes the lamina cribrosa, deep within the optic nerve. Given their lateral size $(2.6 \times 5.2 \text{ mm}^2)$ these images are sufficiently sampled, furthermore the OCT images are better defined than in the previous case since the exposure time is twice as large which improves the SNR.

In Fig. 7(iv) several OCT B-scan slices taken at different $y$ positions are shown with their corresponding SLO image. These feature the optic nerve with a large lateral size of $\sim 5 \times 6 \text{ mm}^2$.

A good match was found between the features visible in the SLO image and those seen in the corresponding OCT B-scans. The correspondence is also clear in (iii) and (v), the former with a significant intercept of a sub-retinal blood vessel and the latter with a good correspondence of features from the foveal region.

5. Discussion

A particular set of three parameters was employed here which allowed simultaneous operation with pixel-to-pixel correspondence: (i) a state of the art line camera performing spectral scanning in 10 $\mu\text{s}$; (ii) lateral scans at 1 ms, the fastest achievable period with a galvo-scanner having a sufficiently large mirror to perform low loss scanning and (iii) a finite bandwidth in the SLO channel, using a 100 kHz APD amplifier. Therefore, the choice of three regimes described here is specific for the current level of Sp-OCT technology combined with that of fast galvo-scanners only and it is not applicable to the swept source-OCT method of spectral domain OCT, where A-scan rates of over 1 MHz are possible [35].

Choosing between the three modes of operation described above, simultaneous, sequential “on-demand” and hybrid sequential, a decision has to be made in terms of the trade-off between the range of configurable parameters and the need of pixel-to-pixel correspondence. Here by parameters we understand the voltage/frequency of the signals driving the lateral scanners and the CMOS camera’s integration time.

The simultaneous mode of operation is suitable for small size imaging only, such as in adaptive optics. If matching the optical resolution with the electronic sampling resolution is not of
Fig. 7. Images obtained with the OCT/SLO set-up operating in hybrid sequential mode (SLO top, OCT bottom). The images in (i), (ii) and (iii) have sufficient sampling whilst the images in (iv) and (v) are under-sampled. (i) and (ii): area between the optic nerve and the shallower retinal tissue (lateral size $2.6 \times 5.2 \text{ mm}^2$); (i): focus on shallow layers; (ii): focus at the lamina cribrosa’s depth; (iii): area located in the vicinity of the optic nerve, the OCT B-scan cut intercepts a blood vessel along its course (yellow arrows), lateral size $2.6 \times 5.2 \text{ mm}^2$; (iv): detail of the optic nerve region ($5 \times 6 \text{ mm}^2$ lateral size) emphasizing several positions of the cursor with varying features selecting the associated OCT B-scans; (v): fovea region, lateral size $5 \times 5 \text{ mm}^2$. The OCT B-scans are obtained from an average of $\Lambda = 4$ OCT frames. The positions of the OCT cuts along the Y-axis correspond to the location of the horizontal lines overlaid on the SLO C-scans.
concern, then the image can be subsampled by increasing the amplitude of voltages applied to the two galvo-scanners. This may be the case when using the SLO channel to perform retinal tracking, when major features in the SLO are sufficient, as suggested in [14].

The sequential “on-demand” mode of operation is suitable for investigation situations where total freedom is needed in terms of image size and quality of image in each channel, irrespective of the quality of signal in the other channel, and when instabilities of the eye position are not of concern.

Lastly, the hybrid mode of operation represents a trade-off between the simultaneous regime and the sequential on demand, where sufficient large size images can be achieved with some degree of pixel-to-pixel correspondence between the images in two channels, moderately affected by the eye movement. This is the regime of operation of the Optos system [12] currently used in clinical investigations.

Different systems exist on the market and in several research labs. Our study suggests that, for instance, if a system is normally being used in the sequential mode of operation, then there is no point in maintaining such mode when the image size is not a determinant factor. Such a system can also, with some modifications as described here, be made to operate in the simultaneous mode, where the frame rate is determined by the speed of the spectrometer and by the speed of the line galvo-scanner in each case.

The notions of pixel-to-pixel correspondence and simultaneity in the simultaneous mode of operation are slightly different from the same notions applied to [Time-domain OCT]/SLO developed in the past [10, 11, 36], where pixel-to-pixel correspondence between the en-face OCT image and the SLO image referred to all pixels in the transversal section of the images. Here, pixel-to-pixel correspondence refers to the lateral cut only, sampled by the B-scan along one of the T-scan lines in the SLO image. It should also be noted that the correction procedure of OCT B-scan image in [37] based on the SLO image collected simultaneously is applicable to the simultaneous mode of operation described here. This is not illustrated here, but with the vertical scanner stopped, lateral movements of the eye can be traced from the breakages of vertical lines in the SLO image, during the acquisition of sets of B-scan OCT images.

As another parallelism of simultaneous technologies, in both TD-OCT/SLO and in the Sp-OCT/SLO presented here, it is the OCT channel that slows down the SLO acquisition. In general, the SLO channel can be made to work faster than the OCT channel. Large size en-face OCT images determine a slower frame rate of the simultaneous acquisition, of 2 − 4 Hz. For reduced image size however, simultaneity was maintained with sufficient signal to noise ratio by using resonant scanners [11]. In Sp-OCT, even though faster frame rates are achievable for the B-scan OCT image in comparison with the ones obtained with en-face TD-OCT, they are not as fast as to allow simultaneity with a large number of pixels in the transversal section; hence the regime with only 100 pixels described here. Replacing the line galvo-scanner with a resonant scanner allows increasing the acquisition speed of the SLO channel, however this will impose a too high speed for the camera. For instance, by swapping the galvo-scanner used here with a resonant scanner at 8 kHz would have meant a 16× faster SLO channel. However, for the same \( N_t = 100 \) pixels, this would have demanded a camera 16 times faster i.e. operating at a 1.6 MHz line rate. The maximum rate the authors are aware of is \( \sim 300 \) kHz [20].

6. Conclusions

In this paper we discussed the possibility of combining a spectrometer based Talbot bands OCT configuration with SLO technology, compatible with limitations imposed by the current technology in terms of speed of linear cameras. While there are transversal scanners which can work faster than the galvo-scanners used in this paper, such as resonant ones, utilization of faster transversal scanners is not compatible with a simultaneous mode of operation OCT/SLO
as described here due to the limited line rates of CCD or CMOS cameras. Utilization of a galvo-
scanner instead of a polygon mirror [38] or of a resonant scanner allows the three different 
modes of operation to be implemented, where the line scanner is slowed down in the OCT 
regime in comparison with the SLO regime to allow for an increased number of A-scans. That 
is not possible with a resonant scanner, which operates at a fixed rate.

Alternatively, the bandwidth of the SLO channel can in principle be increased, allowing more 
 optical pixels to be scanned within the same 1 ms ramp. While a 10 times increase in the band-
width of the SLO channel is possible, this would have ruled out operation in the simultaneous 
mode, as an increase by a factor of 10 of the camera rate (to maintain pixel-to-pixel correspon-
dence) is not feasible. If a faster SLO regime is desired, then the only alternative is to employ 
one of the two sequential regimes presented.

A Talbot bands configuration was employed in this report which allowed the derivation of 
optical signal for the SLO channel with little loss introduced to the OCT channel, however 
the three modes of operation presented can equally be implemented on any other spectrometer 
based OCT channel combined with an SLO channel. A TB configuration in the OCT channel 
redistributes sensitivity from small OPD values to larger OPD values, i.e. a TB configuration 
presents the potential of enhancing the signal from larger depths. In this paper, the sensitivity 
was skewed towards larger OPD values, however the peak sensitivity is less than in a non TB 
configuration, due to the $\text{sinc}$ factor in Eq. (1). The overall sensitivity reduced by $\approx 10 \text{ dB}$ at 
depths below 0.5 mm. This, however, does not present a disadvantage in comparison with con-
ventional Sp-OCT technology. Normally, an imaging depth-range centered on the axial position 
corresponding to $\text{OPD} = 0$ and for a signal roll-off of up to 10 dB is avoided in practice of eye 
 imaging, as a buffer range to cover fluctuations of the axial eye position. Mirror terms might 
occur when due to the axial eye motion, the retinal image is moved to the other side of the zero-
delay ($\text{OPD} = 0$) depth [20]. In other words, the conventional technology presents maximum 
sensitivity within an axial range which cannot be used in practice; more exactly, the sensitivity 
maximum is placed at an axial position in front of the retina. In opposition, a TB configura-
tion can make use of a more suitable sensitivity curve profile versus OPD, with its maximum 
placed within the retina. The larger the gap between the sample and reference beams in their 
way towards the diffraction grating, the larger the shift of the axial position of the maximum 
sensitivity from $\text{OPD} = 0$.

Some loss of sensitivity is also incurred in practice by transferring from a conventional Sp-
OCT configuration to a TB configuration due to the need of securing similar polarization and 
 dispersion in the two beams travelling towards the grating. This problem does not exist in con-
ventional Sp-OCT, as the two beams travel along the same path after they interfered. Therefore, 
more work is required in optimizing the TB configurations to achieve similar efficiencies as 
conventional Sp-OCT configurations. Even so, overall, at larger depths, a TB configuration can 
offer better sensitivity, as proven in [26].

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