HIGH-SPEED *EN-FACE* OPTICAL COHERENCE TOMOGRAPHY SYSTEM FOR CLOSED LOOP ADAPTIVE OPTICS FOR THE RETINA

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Abstract. We previously reported *en-face* Optical Coherence Tomography (OCT) images of the retina using a pair of galvanometer scanners. Unfortunately, the relative low speed of galvanometer scanners has limited the acquisition to 2 frames a second. In order to reach video rate acquisition, we evaluated a resonant scanner at 4 kHz with the aim to replace the galvo-scanner which determined the line in the raster image. In the presented work we evaluated the bandwidth required to generate sufficient lateral size images using such a resonant scanner. The spectrum of the photodetected OCT signal is analyzed depending on the position of the optical beam relative to the axis of rotation of the mirror centered or offset. This system is intended to be used in a adaptive optics closed loop for the retina.

Key words: coherence, tomography, *en-face*, resonant, bandwidth, retina.

INTRODUCTION

In the *en-face* OCT, in order to obtain 3D information about the object, the OCT system is equipped with two orthogonal scanning mirrors, one to scan the object in depth and another one to scan the object transversally. Depending on the order in which these scanners are operated and on the scanning direction associated with the line displayed in the raster of the final image delivered, different possibilities exist. When the two scanners are stopped while the length of the optical path of the reference arm is varied one obtains the depth profile of the reflectivity along a line (this is an A-scan). With one scanner stopped while advancing in depth one obtains a 2D image made of many A-scan lines. This is a B-scan image.

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Usually the transversal scanner produces the fast lines in the image [1, 2, 5, 6]. We call each such image line as a T-scan. This can be produced by controlling either the scanner along the *X*-coordinate, or along the *Y*-coordinate. A T-scan based B-scan is produced, where the *X*-scanner produces the T-scans and the axial scanner advances slowly in depth, along the *Z*-coordinate.

This procedure has a net advantage in comparison with the A-scan based B-scan procedure as it allows production of OCT transverse (*en-face*) images for a fixed reference path, images called C-scans. C-scans are made of many T-scans along *X-Y*, repeated for different values of the other transversal coordinate, *Y-X,* in the same transverse plane. The repetition of T-scans along the other transverse coordinate is performed at a slower rate than that of the T-scans, called the frame rate. In this way, a complete raster is generated. Different transversal slices are collected for different depths *Z,* either by advancing the optical path difference in the OCT in steps after each complete transverse (*X,Y*) scan, or continuously at a much slower speed than the frame rate. For correct sampling in depth of the tissue volume, the speed of advancing in depth should be such that on the duration of the frame the depth variation be no more than half the depth resolution.

Previously we have shown that such *en-face* images can be generated using a galvanometer scanner. Good images from the retina have been obtained operating at 2 frames a second. Higher speed can be achieved using polygon mirrors and resonant scanners. However, this requires an increase of the receiver bandwidth with consequence in the increase in the noise.

MATERIALS AND METHODS

The basic configuration of our experimental set-up is shown in figure 1. Light from a pigtailed superluminiscent diode (SLD, central wavelength λ = 831 nm, spectral bandwidth $\Delta\lambda$ = 17 nm) is split by the beam-splitter BS. Thus, in the target arm, light enters an orthogonal scanning mirror pair consisting from a galvanometer mirror MY and a resonant scanner RS. Both MY and RS scanners could be driven by triangular waveforms at a frequency of 8 Hz and 4 kHz respectively.

After the scanners, light propagates through a telescopic system and enters the eye. Signals from the object and reference arms are directed via microscope objectives into the arms of a single mode directional coupler (DC) to the photodetectors. Finally, the electrical signal was band pass filtered and rectified using a spectrum analyzer (SA).

this Figure is identical with Fig. 1 from ***Bradu2005-2***

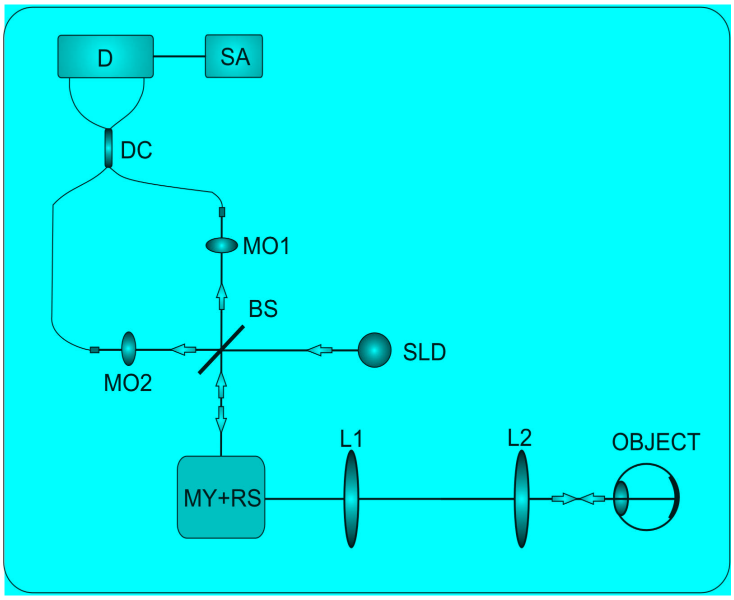

Fig. 1. Block diagram of the *en-face* OCT system; SLD, superluminiscent diode; DC, directional coupler, L1, L2, lenses; BS, beam-splitter, MO1, MO2, microscope objectives; D, detector, SA, spectrum analyser.

The OCT depth could be adjusted via a 1 µm resolution computer controlled stage which changed the optical path of the reference arm. Also, the system could operate in different modes. In the B-scan OCT mode only the galvo-scanner MY was driven with a ramp at 700 Hz and the translation stage (TS) was moved for the depth range required in 0.5 s. In the C-scan mode (*en-face* regime) the MY galvo-scanner was driven with a ramp at 8 Hz and the RS with a ramp at 4 KHz. In order to estimate the bandwidth required, the MY scanner was not driven while various voltages have been applied to the RS (1V, 2V and 3V). For these voltages the estimated images sizes were 1.3, 2.6 and 4.2 mm respectively. Spectra shown in this document have been obtained subtracting from the signal corresponding to a given voltage the signal obtained when the RS is not driven.

RESULTS AND DISCUSSIONS

High image acquisition speeds are required to avoid motion artefacts when retina is imaged. We investigate the effect of a high image acquisition speed due to high scanning frequency on the lateral resolution and signal to noise ratio (SNR) of an OCT system.

There are many factors which the lateral resolution is depending on: optical instrumentation, the limited number of line and frame electronic pixels which can be acquired by the frame grabber, the electronic bandwidth of the receivers, optical aberrations, etc. The limitation of the lateral resolution due to the interface optics of our system was given by the microscope objective used to focus light on the target. We found that this resolution, for the used objective, beam diameter and wavelength was 4.39 µm. When retina is imaged, due to the aberrations, the lateral resolution is limited at about 15 µm [5] depending on the pupil size. As the number of pixels that constitute a line of the image acquired by the frame grabber was 320, the lateral resolution obtained when a line scan of 4.2 mm was 13.12 µm. Thus, if we want the system to operate in a diffraction limited regime the image size has to be decreased. Taking into account the number of pixels in a line scan we evaluated that exploring these pixels at 1.25 ms line scan rate requires a bandwidth of about 0.8 MHz. Considering that the system is diffraction limited, the spot size is 15 µm which determines a lateral size of 1.5 mm. If the image size is increased further, the image bandwidth increases which requires a larger electronic bandwidth. The immediate consequence is a degraded transversal resolution and a smaller value of the signal to noise ratio. Even in these conditions we were able to obtain good *en-face* images of the retina at a frequency of about 8 Hz as shown in figure 2.

The frequency spectrum of the photodetected signal contains components from zero to a maximum frequency that corresponds to a maximum spatial density of the Newton rings. The density of the rings increases from the center towards the edge of the image, so that it looks like the edge of the Newton rings pattern should be used for OCT *en-face* imaging. This is why other techniques as the utilization of an off-axis incident beam or a periodic optical fiber stretching are used.

The consequence of a laterally displacement of the target beam by a certain distance from the axis of rotation of one of the galvo-scanners has as immediate consequence the appearance of the sampling function changes from non equidistant rings to a grid of approximately equidistant and parallels line.

In our manipulations the target beam shifting was introduced by displacing horizontally the beam splitter towards the SLD and then moving the support holding the microscope objective MO1 and the optical fibre.

this Figure is identical with Fig. 2 of ***Bradu2005-2***

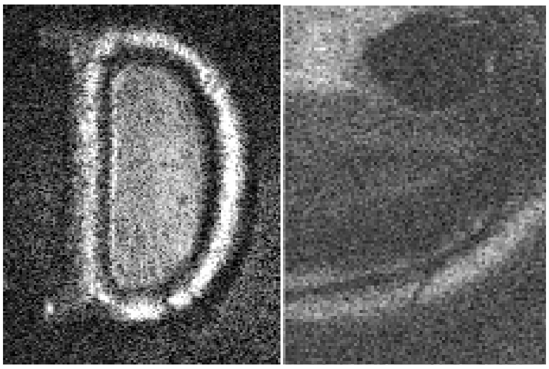

Fig. 2. Examples of *en-face* OCT images obtained with our set-up. The left image represents the letter “D” from a 5 pence coin (1.0 × 1.3 mm) while the right image represents the human retina (2.0 × 2.5 mm).

As it is shown at the left of figure 3 when the target beam is displaced by 3 mm a carrier is distinguishable in the spectrum compared to the non displaced beam spectrum shown at the right of the same figure. For the three situations presented in figure 3 we evaluated displacements of about 1.9, 3.8 and 6.1 MHz respectively which are in good concordance with the experimental signal spectra. Unfortunately, as it can be seen, the beam displacement has as result the enlargement of the OCT signal bandwidth and as a consequence the bandwidth of the photoreceivers and the processing electronics has to be increased, hence a lowering of the signal-to-noise ratio [6]. Thus, for illustration one can evaluate that the bandwidth required to generate a line scan of 4.2 mm in the center case is about 9 MHz, while in the off-axis case of about 12 MHz.

Spectra presented in figure 4 show the evolution of the generated carrier frequency while the beam diameter is diminished from 3.5 mm to about 1.75 mm for different position of the beam on the RS. These spectra correspond to a situation where the image size was 2.6 mm. The first spectrum (top left) corresponds to the centre case for a beam diameter of 3.5 mm. The others situations correspond to a beam diameter of 1.75 mm and various positions Δz of the beam on the scanner.

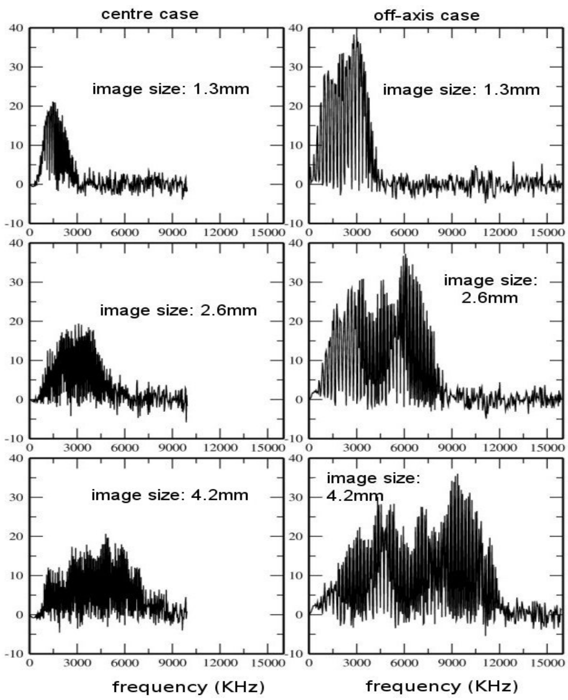

Fig. 3. Frequency spectra of the photodetected signal for the case where the target beam was in the center of the galvo-scanners (left) and displaced by 3 mm (right) for various image sizes.

this Figure is identical with Fig 3 of ***Bradu2005-2***

By analyzing this picture one can conclude that the diminishment of the beam diameter by a factor 2 had as consequence a lowering of the signal bandwidth also by a factor 2 and a displacement of the central frequency of the carrier while the signal bandwidth during the displacement of the beam over the RS remained approximately constant.

Overall, our experiments confirm the fact that by choosing the appropriate values of the beam diameter, the displacement of the beam on the galvo-scanner and electronic bandwidth of the photoreceivers in such a way that one can adapt the carrier frequency and the OCT signal bandwidth to obtain maximum lateral resolution and sharpness of the imaged features in the target.

Even if a good trade-off between the above parameters is done, because of the very large bandwidth of the OCT signal generated by the high acquisition speed the signal to noise ratio will still remain poor. Higher quality images can be obtained only by increasing the optical power. Safety level thresholds for the *en-face* OCT are being evaluated study which shows that larger optical power can be applied to the eye due to a shorter irradiance at each pixel than in OCT cross section imaging.

Nevertheless, when imaging biological samples like retina, the imaged surface are not flat but curved hence an increase of the values of the heterodyne carrier frequencies in the center of the image and a more uniform distribution of the heterodyne frequencies across the image.

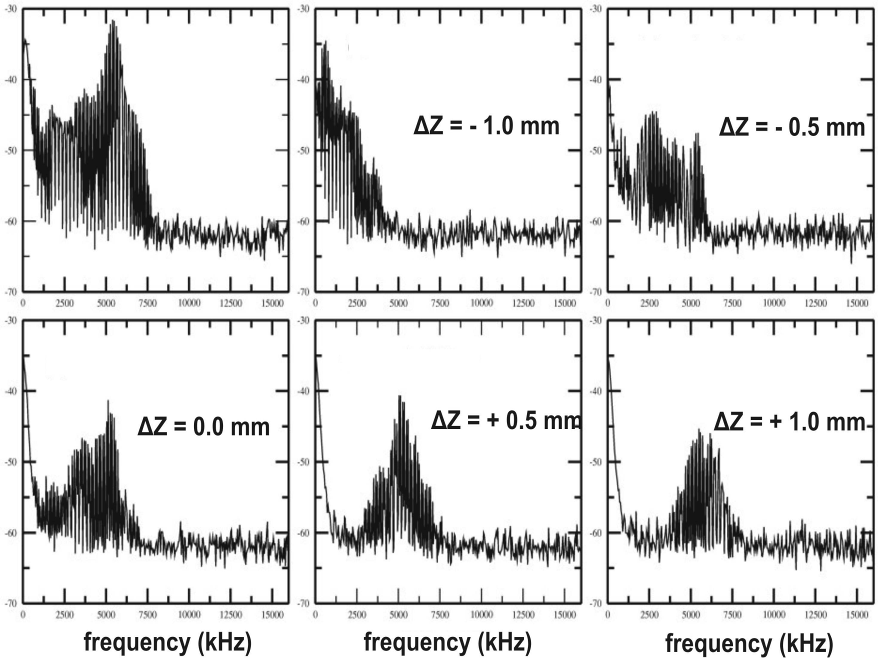

Fig. 4. Frequency spectra of the photodetected signal. The case a corresponds to a situation where the beam was in the center of the RS (Δz = 0) and its diameter 3 mm while the other cases correspond to a beam diameter of 1.75 mm and various position of the beam on the RS.

this Figure is identical with Fig. 5 of ***Bradu2005-2***

CONCLUSIONS

This paper describes the capabilities of a high speed *en-face* OCT system for the retina using a resonant scanner. We evaluated the bandwidth required to generate sufficient lateral size images by analyzing the spectra of the photodetected OCT signals depending on the position of the optical beam relative to the axis of rotation of the resonant scanner centered and offset and on the optical beam diameter.

By means of this system good transversal images from the retina and other samples could be obtained. The system offers speed, required to avoid motion artefacts when retina is imaged, and versatility, being able to display both *en-face* and B-scan OCT images and also confocal images.

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