Covering the gap in depth resolution between OCT and SLO in imaging the retina

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ABSTRACT

Two instruments are now available for high depth resolution imaging of the retina. A scanning laser ophthalmoscope is a confocal instruments which can achieve no more than 0.3 mm depth resolution. A longitudinal OCT instrument uses a superluminiscent diode which determines a depth resolution better than 20 microns. There is a gap in depth resolution between the two technologies. Therefore, different OCT configurations and low coherence sources are investigated to produce a choice of depth resolutions, and to cover the gap between the old confocal technology and the new OCT imaging method. We show that an instrument with adjustable depth resolution is especially useful for the en-face OCT technology. Such an instrument can bring additional benefits to the investigation process, where different requirements must be met. For instance, a poor depth resolution is required in the process of positioning the patient's eye prior to investigation. A good depth resolution is however necessary when imaging small details inside the eye. The utility of the OCT *en-face* imaging with adjustable coherence length for diagnostic is illustrated by images taken from the eye of a volunteer. Images with a similar aspect to those produced by a scanning laser ophthalmoscope can now be obtained in real time using the OCT principle.

Keywords: OCT, low coherence sources, depth resolved imaging, tissue

1. INTRODUCTION

The driving force behind the OCT development for the eye was to improve the depth resolution beyond the limit set by the confocal principle. However, taking into account that the OCT depth resolution can be as low as a few microns, the OCT *en-face* image looks fragmented[1]. Such an image on its own is difficult to interpret. In addition, the shorter the coherence length, the more difficult is to bring the region of interest into coherence. On the contrary, it is relatively easy to bring the image in focus using a scanning laser ophthalmoscope[2] (SLO), where the depth sectioning interval is more than 15 times larger.

Also, with the OCT, the patient head and the eye have to be kept steady within a few micrometers. Micro-saccades of the eye and small head movements disturb the high resolution OCT images more than the SLO images. Consequently, instead of improving the image resolution when reducing the coherence length, we practically end up with deteriorated accuracy.

There is a gap in depth resolution between the two technologies, SLO[2] and OCT[3] when investigating the retina. The optimum depth resolution for retina analysis has not been evaluated so far. A compromise between the increase in the potential depth resolution when increasing the source bandwidth and the disadvantages mentioned above is required. For instance, the ophthalmologists are relatively satisfied by the performances of the SLOs based on confocal principle. An SLO has a depth resolution no better than 0.3 mm and equipped with a powerful software processing tool, gives ophthalmologists access to sub-mm accuracy. Consequently, a natural question arises to what extent it is worth paying the price for applying the OCT technology in a common ophthalmology practice.

Another issue is that the *en-face* OCT image aspect appears different to a confocal SLO image or the image obtained by a fundus camera, again due to the narrow depth sectioning interval of the correlation function used in the OCT. A large data base of images for the diseased eyes exist. The tremendous different aspect of the en-face OCT image (very fragmented) and of a fundus camera or SLO image (quasi-contiguous) makes their comparison difficult. For this goal, it would be desirable to start with a coherence length similar to the depth resolution of the SLO and then reduce it, to make the

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transition from one appearance to the other less abrupt and ease the image interpretation. Unfortunately, the coherence length of a typical low coherent source used in OCT systems, generally SLDs, is not adjustable.

Therefore, we are investigating different procedures to offer the ophthalmologist a choice of depth resolution values, using: (1) a laser diode with a convenient coherence length; (2) a three-electrode laser diode which exhibits spectrum width adjustment under different electric driving conditions; (3) a dispersion controlled OCT set-up when driven by an SLD; (5) software weighted addition of a number of OCT *en-face* images acquired with SLD from adjacent different depths.

Performances of the five methods mentioned above are presented in relation to OCT imaging. Their usefulness for diagnostic is proven by images from the eyes in vivo.

To cover the gap in depth resolution between SLO and OCT, we intend to build OCT systems with a depth resolution of 20 μm to 300 μm. This corresponds to a coherence length of 40 – 600 μm.

2. MULTIMODE LASER DIODES

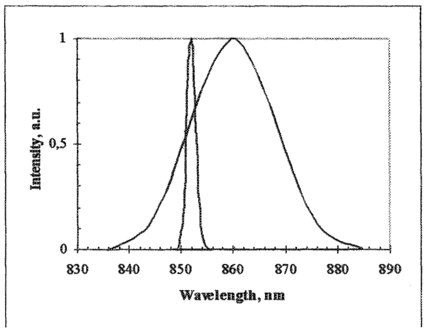

Figure 1 Fig. 1. Normalised emission spectra of the LD ($I_{pumping}$=144 mA) and the SLD.

Commercial multimode laser diodes can in principle be used, however when below threshold, their power in single mode fibre is insufficient to provide a good signal to noise ratio. Therefore, a special laser diode, LD, on 853 nm, was developed by the Technical University of Moldova[4]. This was optimised to deliver at least 0.5 mW in single mode fibre for a current 0.5 mA below the true lasing threshold, when the linewidth becomes too narrow for our application. At this current, a linewidth $\Delta\lambda$ of 2 nm was obtained.

The basic source for our system is a low-coherence single mode fiber pigtailed SLD module with central emission wavelength 860 nm, FWHM-18 nm and 5 mW optical output power. The spectrum of the LD is shown in Fig.1 in comparison with the SLD spectrum. The coherence function for different values of the LD pumping current is presented in Fig.2. The

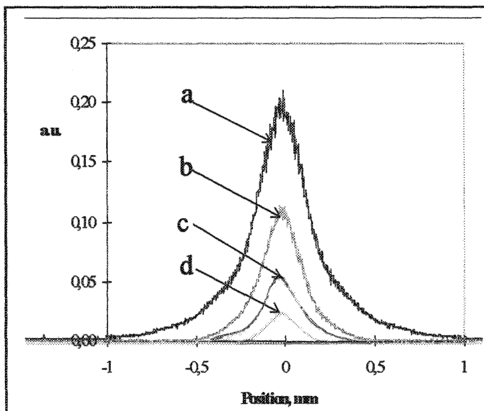

Figure 2. Coherence function envelope of LD 724 (L=1.9 mm) for different values of pumping current: a) I_p=144 mA (l_c=361μm); b) I_p=137 mA (l_c=250μm); c) I_p=129 mA (l_c=222μm); d) I_p=114 mA (l_c=195μm).

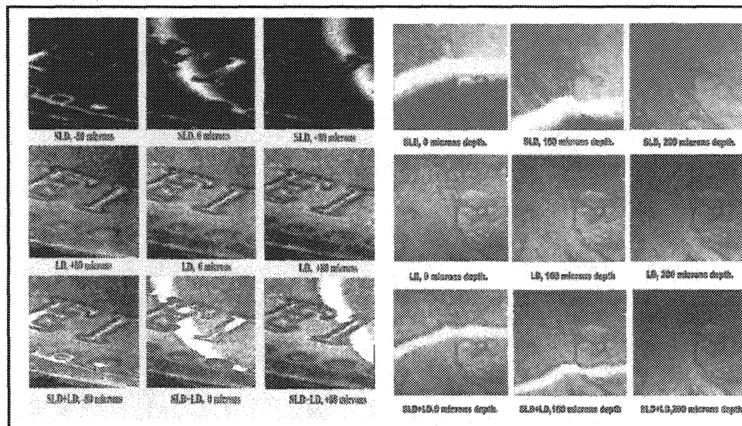

Figure 3. En-face images collected with the LD and the SLD.
The three columns from the left: images from a coin;
The three columns from the right: images from skin tissue in vitro;
Top row: high resolution en-face images collected with the LD only;
Middle row:low resolution en-face images collected with the LD only;
Bottom row: en-face images collected with both SLD and LD sources, equal power from each to the object.
In all cases, the overall power to the object was the same.

coherence length l_c vary from 195 μm to 360 μm for pumping currents between 119 mA and 144 mA according to: