Integrated-optics based multi-beam imaging for speed improvement of OCT systems

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Integrated-optics based multi-beam imaging for speed improvement of OCT systems

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ABSTRACT

The speed improvement is a game-changer in optical coherence tomography (OCT) imaging because it opens up for new and very exciting applications. The frame rate of an OCT system is limited by the speed of the camera or the sweep rate of the light source. This problem can be overcome by multiple-beam imaging, in which different locations on the sample are illuminated by an array of light simultaneously. This technique allows parallel imaging from multiple sample locations and therefore improves OCT axial scan rate by a factor equal to the number of beams used simultaneously which can go up to very high frequency ranges (e.g. MHz) easily. In this work, we introduce a compact integrated-optics based multiple-beam illumination design in which several waveguides with certain length differences are combined with wavelength-independent couplers for space-division multiplexing. Electrodes will be placed on each beam path in order to separate desired signal from unwanted reflections at the optical surfaces or tissue. The imaging speed will be improved by the number of the beam paths used. In addition to fast imaging, the proposed design will be very compact which makes it very suitable to be used in endoscopic probes. The proof-of-concept of this idea was experimentally demonstrated using a design which consists of 2 times 4 parallel OCT channels that are realized with a total of 6 Y-couplers. Each individual OCT channel has an optical path length delay with respect to the other channels.

Keywords: Optical coherence tomography, high-speed imaging, multi-beam imaging, integrated optics

1. INTRODUCTION

Optical coherence tomography (OCT) is a non-invasive optical technique for high-resolution three-dimensional imaging of biological tissues [1]. Current state-of-the-art OCT systems operate in the Fourier-domain, using either a broad-band light source and a spectrometer, known as "spectral-domain OCT (SD-OCT)", or a rapidly tunable laser, known as "swept-source OCT (SS-OCT)". Even though OCT has become a well-established imaging technique in medicine within a very short period of time, there are still some major drawbacks that have to be solved for wider deployment of OCT systems in many field of applications.

One of the main roadblocks is that the imaging speed an OCT system is limited by the speed of the camera (SD-OCT) or the sweep rate of the light source (SS-OCT) which operate at about 20,000 − 300,000 A-scans/s [2, 3]. To overcome this problem parallel detection channels on different OCT configurations have been used by many research groups [4-9]. The parallel imaging technique allows simultaneous imaging from multiple sample locations and therefore improves OCT axial scan rate by a factor equal to the number of beams used simultaneously which can go up to very high frequency ranges (e.g. MHz) easily. Lee *et al.* has introduced interleaved OCT idea in which the full spectral width of the source is divided into P sets of unique spectrally interleaved wavelength components [7]. Each point is illuminated by roughly the full spectral width of the source. However, this system uses complex sample arm design to illuminate multiple locations in the sample (by using virtually imaged phased arrays--VIPA) that lead to around ~10 dB losses in the sample arm. Spectral encoded endoscopy introduced firstly by Tearney *et al.* spectrally spreads the light on to a sample, however it suffers from very poor axial resolution [8, 9]. The existing multi-beam OCT systems have complicated system architecture with multiple interferometers which have prohibited wide utilization of this approach to date.

In this work, an integrated-optics based parallel OCT design is presented. The design consists of several waveguides with certain length differences that are combined with wavelength-independent couplers (i.e. multi-mode interference,

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MMI) for space-division multiplexing. Electrodes will be placed on each beam path in order to separate desired signal from unwanted reflections at the optical surfaces or tissue. The imaging speed will be improved by the number of the beam paths used. In addition to fast imaging, the proposed design will be very compact which makes it very suitable to be used in endoscopic probes. The proof-of-concept of this idea is successfully demonstrated on a device that consists of 2 times 4 parallel OCT channels with a certain optical delay increment between each.

1.1 Parallel OCT design using curved bends

1.1.1 Materials system

The design was made in TriPleX technology platform [10]. The waveguide geometry is a single strip Si_3N_4 of 50 nm height and 3.4 µm width. The top and bottom SiO_2 cladding layers are 8 µm thick. Waveguides operate in single mode at 1550 nm wavelength and have a minimum bending loss for TE polarization.

1.1.2 Device design and experimental demonstration

The chip consists of 2 times 4 parallel OCT channels that are realized with a total of 6 Y-couplers. Each individual OCT channel has an optical path length delay with respect to the other channels, such that they can all be measured simultaneously in one single measurement. The experimental set-up is shown in Fig. 1 and the chip layout is shown in Fig. 2(a). Figure 2(b) shows an image of the measured chip with a 635-nm laser coupled at the input showing all 8 OCT channels.

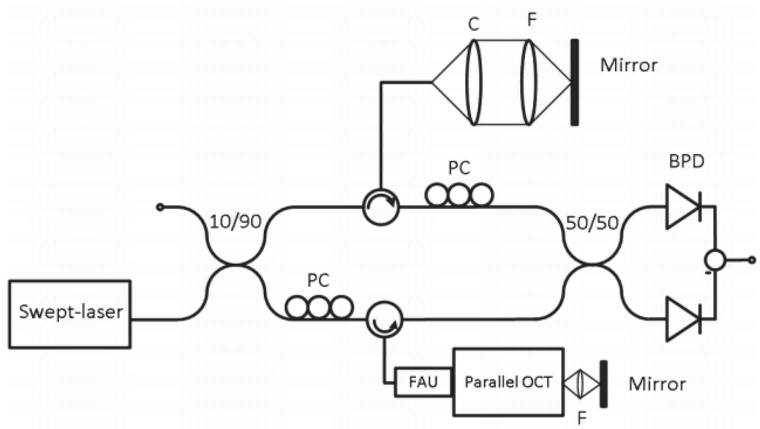

Figure 1. Experimental set-up to characterize the parallel OCT chip: BPD: balanced photodetector, FBG: fiber Bragg grating, PC: polarization controllers, C: collimating lens, F: focusing lens, and FAU: fiber array unit.

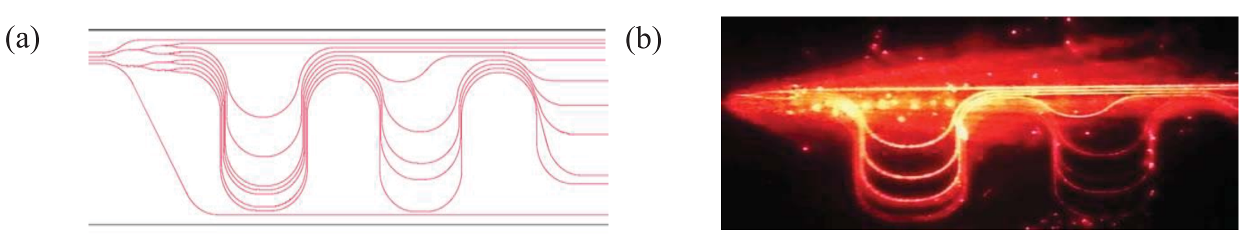

Figure 2. (a) Layout of the parallel OCT sample arm chip with 8 parallel channels. (b) Image of the chip with light coupled from 635 nm laser.

For the purpose of characterizing the chip, 4 channels were sequentially measured by translating the lens from channel #1 to channel #4. In principle, with the choice of an appropriate imaging lens, all 8 channels can be measured in parallel. Figure 3 shows plots composed of 4 sequentially measured OCT channels. During the measurement, the position of the sample and reference arm mirrors was not changed. The optical path length difference shown in Fig. 3(b) is created by the on-chip length difference. The measured delays lines are 2.22 mm, 4.62 mm, 7.27 mm, which are in good agreement with the designed delay lines of 2.5 mm, 5.0 mm, and 7.5 mm.

Due to the material refractive index mismatch between the sample arm (TriPleX) and the reference arm the width of the signal corresponding to the position of the mirror is broadened. To return the signal to its bandwidth limited width we applied dispersion compensation based on a numerical algorithm.

The power budget for this chip is as follows. The input power for the fiber-array unit was 6.1 mW. The measured transmission power for the alignment channel was 937 μW. The measured transmission powers at the 4 channel are 153 μW, 195 μW, 226 μW, and 223 μW, respectively.

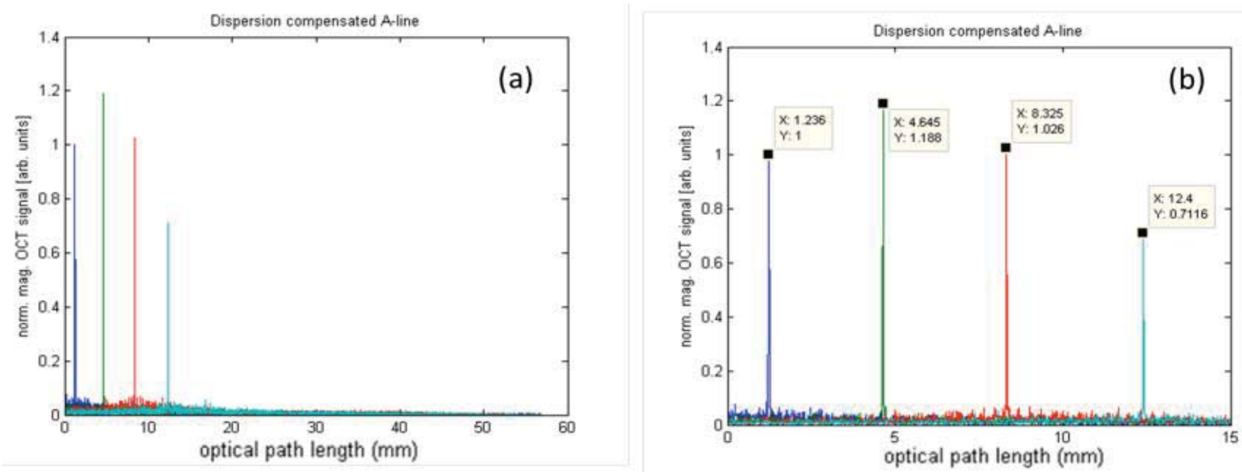

Figure 3. (a) A-lines for the first 4 OCT channels of the parallel chip. (b) The measured delay lines are 2.22 mm, 4.62 mm, 7.27 mm.

1.2 Parallel OCT design using curved waveguides with branches

An illustration of the sample arm configuration with the multi-beam imaging design using curved waveguides in combination with branches is shown in Fig. 4. Input light will be divided into equal-power arms by using Y splitters. In the end of each waveguide the light will be divided into three branches using multi-mode interference couplers (MMI) with equal power. Each branch has a different optical delay. For the example considered here, a total of 12 different beam paths are demonstrated; 4 main arms with 3 branches in each. This allows the benefits of being able to use more power to illuminate the sample safely. The length differences between arms are arranged in such a way that OCT peaks coming from each sample location will be apart from each other by depth range (i.e. x in here, see Fig. 4 (b)) in the Fourier domain. Different illuminated points on the sample have different optical path lengths and the interferometric signal from each point will be extracted from a different imaging depth. The branches at the end of each beam path will have the same design in each arm, so the optical delay between each path will be provided by the main arms. Electrodes placed on each beam path will be used to separate desired signal from unwanted reflections at the optical surfaces or tissue which is one of the biggest problems in multi-beam imaging techniques. Light coming from each arm can be focused into tissue using a large-size focusing lens or single integrated-optics focusing lenses [11] at the end of each branch. In order to reduce the size of the device, spirals can be used instead of curved waveguides in the main beam paths. Assuming a lateral resolution of 20 μm, 50 branches and thereby 17 main arms will be needed for a scanning range of 1 mm. The maximum delay line will be around 2.6cm which can be designed in a very compact way by using spirals. The expected overall sample arm size will be around 1.7 mm × 2 cm. The width of the sample arm can be reduced further by using a high index material system (e.g. silicon) with smaller bending radius in expense of length of the sample arm. Since the most critical dimension in endoscopic applications is the width of the inner circuit, the proposed design can still be quite attractive for forward-looking endoscopic probes despite its long length. In summary, this technique enables synchronized cross-sectional and 3D imaging of biological samples, and provides a new cost favorable approach to further improve OCT axial scan rate.

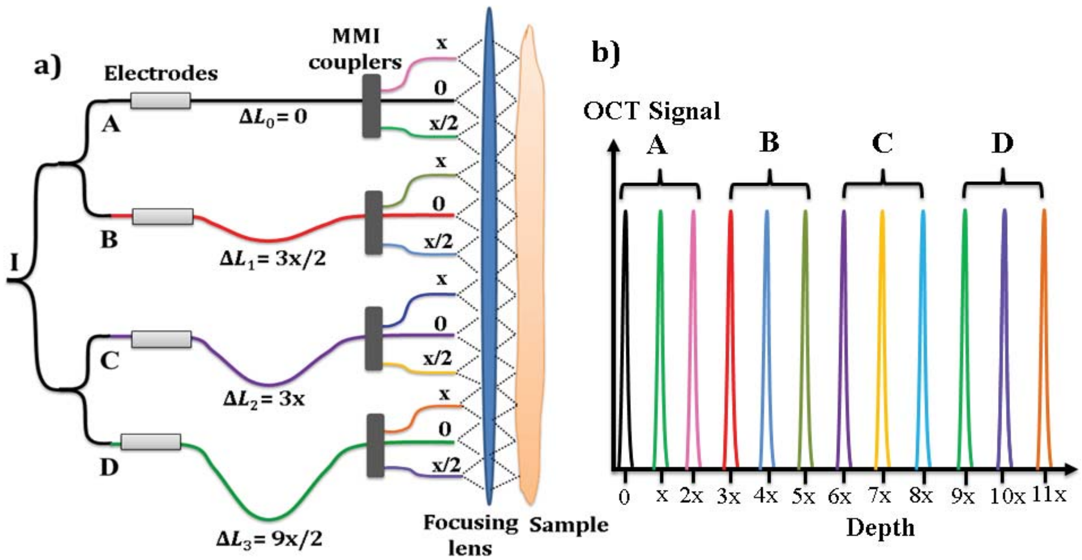

Figure 4. a) Schematic of the proposed parallel OCT system. Input light is divided into several arms with equal power using Y splitters. Each arm then divided into three sub-branches using multi-mode interference couplers (MMIs). Here in total 12 optical beams will be generated. The path length difference between sub-branches are the same for each arm (i.e. 0, x/2, and x) whereas there is a 3x/2 difference between each main arm. Using a large focusing lens each beam can be successfully focused into tissue and the scattered light can be collected back in the same way. b) OCT peaks obtained by using parallel OCT design. Each color represents an individual imaging location as indicated in a).

CONCLUSIONS

In summary, integrated-optics based parallel OCT system designs were presented. The experimental demonstration of the first design which is based on curved waveguides was given. As a more compact alternative to the curved waveguides, a design with sub-branches with certain length difference between each branch was proposed. The experimental demonstration of the alternative design will be given in the future work.

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