L.8: Bidirectional single detector based spectral domain Doppler optical coherence tomography

Doppler optical coherence tomography (DOCT) is an emerging high spatial resolution technique for non-invasive imaging of blood flow in tissue microvasculature. The ability of DOCT for simultaneous evaluation of tissue morphology and blood velocimetry can be of considerable diagnostic help in diabetic retinopathy, glaucoma as well as management of wounds. Doppler OCT detects change in the frequency of the scattered light for quantification of blood flow in tissue microvasculature. The Doppler measurement requires prior knowledge of Doppler angle (i.e. the angle between direction of incident beam and flow) to get absolute velocity information. However, this is most often not possible in biological tissues. In samples having major blood flow confined to a given plane, velocity can be measured by simultaneously probing the same spatial location in the sample with two beams with different incidence angles in a plane parallel to the flow direction.

A major shortcoming with angularly displaced two probe beams is that in addition to two images corresponding to the two probe beams retracing their original paths after scattering from the tissue, a third image is also generated. This corresponds to the first probe beam being scattered along the path of the second probe beam and vice-versa. Because of this additional 'cross talk' image the depth of the tissue that can be imaged with a given setup gets reduced. This is more serious issue with Fourier Domain OCT (FDOCT) systems where the imaging range of the setup is limited by the spectral resolution of the spectrometer. Furthermore, in FDOCT systems the reduced sensitivity with increasing depth of imaging degrades phase measurement accuracy and this eventually affects absolute velocity estimation.

The undesired cross talk image can be removed by the use of two separate interferometers and detection units that share the same sample and reference arms but have an optical delay longer than the imaging range of the systems. However the use of two separate interferometers and spectrometers makes the system costly and bulky.

At Laser Biomedical Applications and Instrumentation Division, we have developed a simpler, approach to address this issue. This involves incorporation of a beam displacer (BD) in the sample arm of the fiber optic interferometer generates two orthogonal polarization beams. Since the BD is bidirectional in nature, the light scattered from the sample that retraces its incidence path, only pass through aperture A1 placed before the BD and recombine. The scattered light following cross-path will get blocked by the aperture A1. The light beams reflected from the reference and sample arms recombine at the detector and generate interference fringes. These delay encoded interference patterns are detected by spectrometer equipped with a line scan camera. The acquired interference spectra are resampled to generate evenly spaced interference spectra in k-space. The FFT of the resampled interference spectrum generates the depth resolved amplitude (structural) and phase images. By making use of the phase difference between adjacent A-scans in each of these phase images, the flow velocity component along their respective beam directions can be determined.

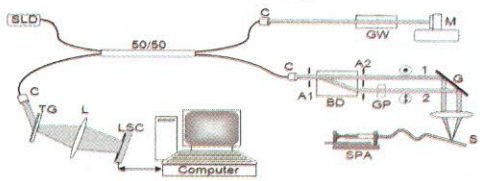

Fig.L.8.1: Schematic of the developed dual beam spectral domain Doppler Optical Coherence Tomography setup. A1,A2-apertures, BD-beam displacer, C-collimeter, G-galvanoscanner, GW-glass window, L-lens, LSC-Line scan camera, M-mirror, SLD-superluminiscent diode, S-syringe, SPA-syringe pump assembly, TG- transmission grating.

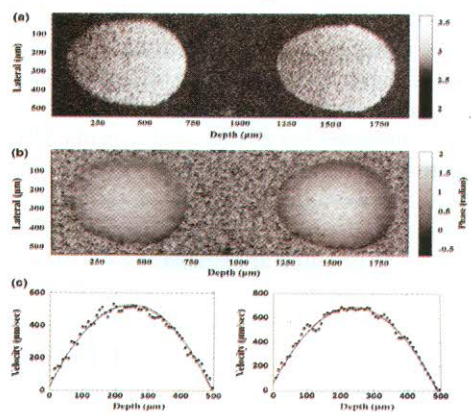

Fig.L.8.2: (a) The structural images corresponding to the two probe beams. (b) The phase-resolved DOCT images corresponding to the two beams. (c) Velocity distributions along each of them. The points represent the experimentally obtained data while the solid curves represent fitted fit to the data.

The setup has been validated using a syringe pump assembly and 1% Intralipid solution in water flowing through a glass tube with 0.5 mm inner diameter. Fig. L.8.2a and 2b show the structural and phase images of the intralipid flow phantom formed by the two orthogonal polarization beams. Velocity components v1 and v2 were calculated by parabolic fitting of the measured velocity profiles. These are used to calculate Doppler angles and the velocity (V) in the plane containing two probe beams. The measured axial velocity distribution profiles for each beam are shown in Fig. L.8.2c. The estimated velocity values are found to be in good agreement with the set values. The average root mean square (rms) error in the velocity measurement was ~0.05 mm/s. For more details, please refer to S. Kumar et al., Applied Physics B : Lasers and Optics, 2014 (DOI: 10.1007/s00340-014-5848-4)

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