Compensation of motion artifacts in catheter-based optical frequency domain imaging

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Abstract: A novel heterodyne Doppler interferometer method for compensating motion artifacts caused by cardiac motion in intracoronary optical frequency domain imaging (OFDI) is demonstrated. To track the relative motion of a catheter with regard to the vessel, a motion tracking system is incorporated with a standard OFDI system by using wavelength division multiplexing (WDM) techniques. Without affecting the imaging beam, dual WDM monochromatic beams are utilized for tracking the relative radial and longitudinal velocities of a catheter-based fiber probe. Our results demonstrate that tracking instantaneous velocity can be used to compensate for distortion in the images due to motion artifacts, thus leading to accurate reconstruction and volumetric measurements with catheter-based imaging.

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1. Introduction

The integration of Fourier-domain detection strategies in optical coherence tomography has opened the possibility of high-resolution, cross-sectional imaging over large fields of view in biological tissues [1–4]. Optical frequency domain imaging (OFDI), which is based on frequency domain ranging techniques, has been applied through narrow diameter catheters and endoscopes [5] for imaging the coronary arteries [6–9] and distal esophagus [10] in patients. In intracoronary OFDI, a chirped beam probes the radial profile of the vessel being imaged. The probe collects the scattered light and transmits it to the proximal end of the catheter, where the scattered light interferes with a reference beam to produce a spectral fringe pattern that mimics the radial profile of the vessel wall. OFDI side-imaging probes consist of single mode fiber and distal optics with a microlens and a prism, or a polished ball lens fused at the fiber tip. The axial resolution, which is about 10 µm, depends on the wavelength sweep range of the imaging beam and the transverse resolution is determined by the spot size. As most probes incorporate small lenses for focusing the imaging beam, the transverse resolution is on the order of 10 to 20 µm.

In order to create a two- or three-dimensional map of vessels, the fiber probe is inserted into a vessel through a transparent stationary outer sheath and rotated with an automated pullback as shown in Fig. 1. Here, a rotary junction motor spins the fiber in its sheath at a rate of about 100 revolutions per second, while a linear stepper motor pulls the fiber back within the sheath at speeds of 5 to 20 mm/s [6]. Assembling cross-sectional images acquired during the pullback movement typically assumes that the measured imaging data are spaced at equal intervals along the helical track. Although the motor controls the velocity of the fiber probe relative to its protective sheath, the sheath itself can move significantly within the coronary artery during the cardiac cycle, resulting in image distortion (Fig. 2). Although distortions arising from displacements of the sheath that are perpendicular to its axis can be partially compensated through simple surface-aligning algorithms [5,6], these routines artificially straighten the vessel and, as a result, data processed in this manner lose the true, natural contour of the coronary wall. Accurately correcting longitudinal displacements based on image characteristics alone is even more challenging.

In this paper, we demonstrate a novel motion tracking system using a heterodyne Doppler interferometer to compensate for image distortion in intracoronary OFDI. This method uses dual, wavelength division multiplexed (WDM) monochromatic beams to track the relative radial and longitudinal vessel motion. Motion of the vessel is acquired by a frequency shift of the backscattered light caused by the Doppler effect. The re-registration of scans is performed with the Doppler frequency information, showing the improvement in endoscopic images.
Fig. 1. Intracoronary OFDI method and image data displacement due to vessel motion.

Fig. 2. Longitudinal reconstruction of a 3D intracoronary OFDI data set obtained from a swine coronary artery in vivo. Yellow arrows mark locations of significant motion artifact caused by cardiac motion.

2. Principles

2.1 Doppler tracking

To create a cross-sectional image of the vessel, the radial profile readings at each spot in the helix are assembled. The information of the radial displacement and longitudinal displacement is required to correct displaced image data. Figure 3 schematically depicts the configuration of the distal optical end of the catheter probe, including the dual WDM tracking beams. To track the radial and longitudinal probe motion with regard to the vessel, the radial and longitudinal monochromatic tracking beams illuminate on the vessel wall at a different incidence angle. The instantaneous probe motion relative to the vessel wall is then measured by an amount of Doppler shift which depends on the relative velocity, the incidence angle, $\theta_i$, and the tracking wavelength, $\lambda$:

$$f_D = \frac{2}{\lambda} (\overline{v_r} \cos \theta_i + \overline{v_z} \sin \theta_i)$$  \hspace{1cm} (1)$$

where $\overline{v_r}$ and $\overline{v_z}$ are the relative radial and longitudinal velocities. Since the dual beams illuminate at different angles, they experience different Doppler shifts for the same velocities. As a result, the velocities are measured in terms of the known incidence angles and the measured Doppler shifts:

$$\begin{pmatrix} \overline{v_r} \\ \overline{v_z} \end{pmatrix} = \frac{1}{2 \sin (\theta_i + \theta_2)} \begin{pmatrix} -\sin \theta_2 & -\sin \theta_1 \\ \cos \theta_2 & -\cos \theta_1 \end{pmatrix} \begin{pmatrix} \lambda_1 f_1 \\ \lambda_2 f_2 \end{pmatrix}$$  \hspace{1cm} (2)$$
2.2 Heterodyne Doppler beating

Measurements of Doppler frequency shifts are achieved by mixing backscattered light from the vessel wall with light from the heterodyne reference beam, also known as a local oscillator (LO). The detected signal associated with interference is approximately given by:

$$I(t) \approx I_{LO} + \sqrt{I_{LO} I_1} \cos[2\pi(f_{LO} - f_1) t] + \sqrt{I_{LO} I_2} \cos[2\pi(f_{LO} - f_2) t]$$

where $I_{LO}$ is the heterodyne reference intensity, $f_{LO}$ is the LO frequency, and $I_1$ and $I_2$ are the intensities of the returned radial and longitudinal tracking beams. Doppler frequency shifts, $f_1$ and $f_2$, change as the vessel moves. Information of the frequency of the LO is utilized to distinguish the direction of vessel motion and to determine the corresponding relative velocity through Eq. (2). The heterodyne reference beam is used for amplifying the backscattered signal and distinguishing the direction of motion. For instance, when $f_{LO} = 0$, equal but oppositely directed velocities produces signals that oscillate at the same frequency because of the directional ambiguities of the cosine. When $f_{LO} > f_1$ and $f_2$, equal but oppositely directed velocities cause the detected signal to shift away from the LO in opposite directions, removing this ambiguity. To generate a LO, a frequency shift can be induced in the reference arm using an acoustic-optic frequency shifter. It should be noted that the use of phase modulators or piezoelectric actuators is not applicable due to the generation of two heterodyne reference signals (up and down frequency-shifted signals) relative to the original source signal which are mixed with a backscattered Doppler signal together. As a result, the Doppler beatings are generated at the double side bands relative to the heterodyne reference at $f_{LO}$ which cannot provide the direction of motion.

2.3 Data processing and image compensation method

In catheter-based rotational imaging, the spatial coordinates are relative as the first pixel determines the origin of the image coordinate system. Thus, tracking the velocity instead of the position can provide enough information to compensate for motion induced distortion. Here, the velocity is encoded in the beat frequency of a Doppler interference signal, which is recorded during the entire image data acquisition. Analyzing the time series Doppler data using any one of time-frequency distributions yields the instantaneous Doppler frequency shift, which, in turn, is proportional to the velocity. Converting the frequency to velocity and multiplying the result by the time interval gives an estimate of the actual displacement between data samples. In this work, the most commonly used short time-frequency transform (STFT) is employed [11]. The time series data are divided into small time segments and a Fourier transform is applied to each segment to extract the frequencies in the finite sampling
time, which produces a trade-off between time resolution and frequency resolution. The width of the STFT window determines the precision of the velocity measurement. The Fourier transformed data represents the power spectrum density with regard to time and frequency simultaneously.

The compensation procedure is performed with the intensity based image registration. The OFDI data set, \( I(r(t), z(t)) \), is re-registered by the recorded Doppler displacement data, \( [r'(t), z'(t)] \). To correct the radial motion artifacts, the matrix data sets are simply shifted by the amount of displacement. Special care is required for the longitudinal correction as illustrated in Fig. 4. Without motion artifacts, the longitudinal data points are acquired with equal intervals by the constant probe pullback motion. However, the overlapped data points are generated with an abrupt reverse pullback motion (-z direction), generating a negative displacement. Instead of removing the overlapped data, the image data sets are re-registered by changing intensity values. For instance, the successive image data of the same position are generated with the same intensity when the probe motion is stationary. To correct the image, the intensity of the first image data is simply divided by the number of data points generated during the image acquisition. For small displacements between data points, the intensities of adjacent image data are changed depending on the amount of the displacement from the adjacent data points. When an abrupt positive displacement (+ + z direction) generates an image gap, it is filled with interpolated data between the adjacent data points.

![Distortion compensation method](image)

**Fig. 4.** Distortion compensation method.

### 3. Experiments and discussion

#### 3.1 Sensitivity analysis

To analyze the performance of the proposed tracking system, a theoretical sensitivity analysis is presented. The signal-to-noise ratio (SNR) is the ratio of the signal power to all the noise contributions. These considerations in a tracking system are similar to those associated with OFDI systems [4,12,13]. Apart from an OFDI system the additional effect of phase noise should be considered in a tracking system due to the use of a long coherence length laser source such as a distributed feedback (DFB) laser [14]. Phase noise is caused by finite laser linewidth or variations of optical delay between a reference signal and a sample signal. The effect of phase noise can be minimized by choosing a laser with a narrow linewidth or matching path lengths between two arms. Although the path length is matched for the
backscattered Doppler signal from the vessel, there still exist stationary heterodyne beating signals which are generated by the crosstalk signal of the circulator and the back-reflections from the facet of the distal catheter optics, the fiber connections, and the body of components. These path-length mismatched signals contribute to the Doppler beating signals as background noise. The only solution to avoiding this problem is to create a large enough frequency difference between the desired and the heterodyne beat frequencies. However, the Doppler frequencies generated by vessel motion in intracoronary OFDI are ranged within the phase noise affecting region. Therefore, the total noise contributions of a tracking system should include the phase noise as well as the detector thermal noise, shot noise, and relative intensity noise (RIN) of the laser source. The signal power in a dual balanced detector can be expressed by

\[
\langle i^2_{\text{signal}}(t) \rangle = 8 \left( \frac{\eta q}{hv} \right)^2 p_r p_s \tag{5}
\]

where \( q \) is the electric charge, \( \eta \) the quantum efficiency of the detector, \( h \) is Plank’s constant, \( v \) the frequency of the laser source, and \( p_r = P_r/2 \) and \( p_s = P_s/2 \) denote the reference and signal power per photodiode. The noise power including the phase noise term can be given by

\[
\langle i^2_{\text{noise}}(t) \rangle = \langle i^2_n \rangle + 2B \left( \frac{\eta q}{hv} \right)^2 \sum p_r + p_s + \left( \frac{\eta q}{hv} \right) RIN \left[ \sum \zeta (p_r^2 + p_s^2) + \sum 2p_r p_s \right] \\
+ B \left( \frac{\eta q}{hv} \right)^2 RIN_{\text{LO}} \left[ \sum \zeta (p_r^2 + p_{\text{talk}}^2) + \sum 2p_r p_{\text{talk}} \right], RIN_{\text{LO}} \equiv \frac{2p_r p_{\text{talk}}}{p_r^2 + p_{\text{talk}}^2} 8\pi \tau_0^2 \Delta v \sin^2(\tau_0 f_{\text{LO}}) \tag{6}
\]

where the four terms on the right hand side represent the thermal noise, shot noise, RIN noise, and phase noise caused by the crosstalk signal of the optical circulator, respectively [15]. \( \zeta \) is the common-mode rejection efficiency of the balanced receiver, \( \tau_0 \) the differential delay for the crosstalk signal of the circulator, \( \Delta v \) the linewidth of the DFB laser, \( B \) the electric FFT spectrum bandwidth, and \( P_{\text{talk}} \) the power of crosstalk signal per photodiode. In this calculation, the DAQ noise is ignored due to very small contribution to the total noise and the following parameters of the system specifications were used: \( p_r = 226 \text{ nW}, P_{\text{talk}} = 7.94 \text{ nW}, \)

\( i_{\text{th}} = 2.5 \text{ pA/sqrt(Hz)}, \eta = 1, f_{\text{LO}} = 1 \text{ MHz}, RIN = -155 \text{ dB}, \zeta = -25 \text{ dB, and } B = 5 \text{ MHz.} \)

Fig. 5. System configuration for measuring sensitivity.

We measured the sensitivity of the tracking system as a function of reference arm power for a single reflector as shown in Fig. 5. The DFB laser operating at a wavelength of 1550 nm was used to generate the tracking beam. 1% of the WDM coupled light was directed to acousto-optic modulators (AOMs), while 99% was directed to the sample arm. In order to
distinguish the direction of motion, a heterodyne reference signal at a frequency of 1 MHz was generated by using two cascaded AOMs (Brimrose, Inc.) at a driving frequency of 25 MHz and 24 MHz, respectively. A frequency shift of 1 MHz relative to the optical laser frequency was high enough to accommodate the bandwidth of the Doppler signal and low enough to be sampled with a DAQ board. The neutral-density (ND) filter with an insertion loss of 41 dB was positioned to control the sample arm power. The path length between the two arms was matched for the back-reflection from a metal mirror. To evaluate the effects of phase noise on Doppler beating signals, the sensitivity was measured in the phase noise dominant region and non-dominant region away from the crosstalk beating frequencies. The sensitivity measurement procedure is illustrated in Fig. 6. First, the background noise was measured without sample motion. Next, a Doppler signal was generated with an oscillating mirror. Finally, the signal power and the saved background noise power were compared and then the insertion loss of the ND filter was added. Figure 7 shows the theoretical SNR curves and the measured SNR as a function of reference arm power. The SNR was also measured when the different path delayed crosstalk beating exists. Here, the contribution of back-reflections from the ND filter and collimator to phase noise were ignored due to their nearly matched path length. The results show that the measured data are well matched with the theoretical values and the sensitivity is degraded as the unbalanced path length of a crosstalk signal increases. The measurement error also becomes larger because of the enhanced randomness of the phase noise.

![Fig. 6. Sensitivity measurements (a) with phase noise and (b) without phase noise.](image)

![Fig. 7. Theoretical and experimental sensitivity analysis.](image)
3.2 Integration of motion tracking system with OFDI

Figure 8 shows the experimental setup for demonstrating Doppler motion tracking. The tracking system was combined with a standard OFDI system by using WDM techniques. The wavelength-swept laser using a semiconductor optical amplifier and polygon scanning filter was operated at a sweep repetition rate of 51.4 kHz with a wavelength sweep range of 120 nm. The OFDI system provided an axial resolution of ~7 µm and a transverse resolution of 30 µm with a ranging depth of 6.3 mm. Two DFB lasers, having wavelengths of 1550 nm and 1603 nm, were incorporated into the tracking system to register radial and longitudinal motion. To test the feasibility of Doppler-based velocity tracking without requiring the development of a rotary junction capable of transmitting the OFDI wavelengths as well as the DFB laser wavelengths, the apparatus was simplified to use a rotating phantom. The phantom comprised a 3 mm diameter nylon tube with perforations through one-wall and into the adjacent wall. Images of the phantom were acquired while rotating it at 50 revolutions per second and translating longitudinally with a pullback velocity of 2.5 mm/s. Nonuniform motion was simulated by independently driving the probe using a galvanometer. Figure 9 indicates the spatial orientation of the two tracking beams. At the distal end of the probe, the radial tracking beam at a wavelength of 1550 nm along with the OFDI imaging beam operating at a center wavelength of 1310 nm reflected from the facet of a polished ball lens and illuminated at an incidence angle of 10°. The longitudinal tracking beam, at a wavelength of 1603 nm, reflected from a metal mirror through a GRIN lens with a focal length of 2 mm and illuminated the nylon wall at an incidence angle of 49°. The illuminating power of the radial and longitudinal tracking beams were 7.5 and 5.5 dBm, respectively.

![Diagram of motion tracking system](image)

Fig. 8. Combination of a tracking system with a standard OFDI system. LD: laser diode; WDM: wavelength division multiplexer.
3.3 Doppler frequency measurement and image compensation

To evaluate Doppler-based velocity measurement, both radial and longitudinal Doppler frequency shifts were measured as shown in Fig. 10. The galvanometer scanner supporting the probe was driven with a 60 Hz sine wave and roughly aligned to generate motion in both the radial and longitudinal directions. As can be seen in the data of Fig. 10a, the resulting Doppler shift is registered in both channels with different magnitude and direction. The interference signal with a constant 1.0 MHz tone resulted from crosstalk in the circulator. Additional modulations at the same frequency but with smaller amplitude and sign were observed in the Doppler frequency plot for the longitudinal beam because of the small crosstalk signal of the radial tracking beam at 1550 nm. The velocities of the two motions through Eq. (2) were calculated and are displayed in Fig. 10b.

In order to simulate nonuniform motion of a vessel, the fiber probe attached on the galvanometer was moved back and forth with 3 Hz triangle waves, which achieved velocities up to ~100 mm/s. OFDI image data consisted of 1024 × 1024 × N (N = number of frames) pixel data sets, which were processed using ImageJ to display images of the internal surface of the phantom as if the cylinder was unwrapped [16], as well as longitudinal cross-sections.
through the region of two partial-thickness holes opening from the lumen. To perform a frequency analysis of the Doppler data, an STFT window length of $2^{11}$ was used. This limited a minimum resolvable velocity of 5.2 mm/s for the longitudinal tracking beam. Imaging was first performed with uniform scanning only, simulating the case free from image distortion (Fig. 11a). Subsequently, the galvanometer was used to simulate uncontrolled sample motion. In this case, OFDI data demonstrated significant distortions as depicted in Fig. 11b. Most significantly, the structure of the holes within the phantom lumen appeared distorted and there was multiple replication of individual holes. By acquiring the Doppler tracking signals simultaneously with the OFDI image data and applying the distortion correction procedure outlined above, however, structural fidelity was improved (Fig. 11c). Although the performance of the image compensation was limited by the DAQ acquisition speed of 10 MS/s, our results demonstrate that the motion compensation technique significantly improves the longitudinal view of the phantom.

Fig. 11. Longitudinal views (a) without probe motion artifacts and (b) before and (c) after distortion compensation.

4. Conclusion

We have proposed a novel motion tracking system using a heterodyne Doppler interferometer to monitor relative displacement between an OFDI imaging probe and a phantom. Our proof-of-principle experiments demonstrate that tracking actual instantaneous relative velocity of a catheter can be used to correct for distortion in the images captured during endoscopy. Although the performance of this approach may be improved using faster digital acquisition hardware, our results indicate that even an inexpensive 10 MS/s board provides adequate compensation for motion characteristic of that experienced during intracoronary imaging.

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