Efficient Signal Processing Algorithm for Fast Optical Doppler Tomography Systems

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Abstract – *Optical Doppler tomography (ODT) is an imaging technique which uses the Doppler shift observed in the spectrum of the backscattered light to form images of blood flow in a sample tissue. As data acquisition rates for ODT systems become greater, the need arises for a fast algorithm able to deal with increasingly short time records of the backscattered light. The objective of this paper is to present a potentially real-time algorithm that is an accurate estimate of the flow vector and is robust to noise. Our contribution is that we transform the physical model of the data, stemming from the nature of ODT, into an AM-FM model. The proposed estimator, rooted in the AM-FM model, finds the Doppler shift from a phase shift. The phase shift is caused by the Doppler shift in a small time interval between the recording of the signal and its later version. The calculated Doppler shift is used to form the image of the blood flow in the tissue.*

I. Introduction

Optical Doppler tomography (ODT) is a non-invasive imaging technique, which incorporates laser Doppler flowmetry and optical coherence tomography to produce images of static and dynamic components in highly-scattering biological samples. In ODT, light is emitted from a partially coherent source and the intensity of the backscattered light is measured. Any flow in the tissue that is in line with the source beam will result in backscattered light whose spectrum exhibits Doppler spread (σ_d^2) and shift (f_d) away from the single carrier frequency (f_c) source. Doppler shift is proportional to the velocity of the blood. Combination of the reflected signal and the reference signal produces an interference fringe pattern (IFP). This IFP contains the information about the structure and blood flow in the tissue. ODT has the ability to image the biological tissue in all three spatial dimensions. Since the lateral movement of the source beam can be controlled very precisely (2-15 μm), high spatial resolution images are obtained.

In Section II we will briefly review the short history of this imaging technique and current data processing methods used in ODT. Section III will explain why a new algorithm is needed for processing ODT signals. The mathematical formulation of the problem will be laid out in Section IV. Section V covers the steps for our proposed algorithm. In Section VI, we derive the performance limitations of our algorithm, and in Section VII, we show the results of our algorithm run on simulated data. Finally in Section VIII, we summarize the contributions of the project.

II. Background

After the invention of the laser in the late 1960's, the first references to ODT appeared in the early 1970's [1]. However, for many years, this method of imaging was plagued by low-resolution and inaccuracy. In 1995, a completely new method of ODT was developed in [2], which solved many of the problems with earlier systems. The two notably active research groups in this area are at the University of California at Irvine [2, 3] and Case Western Reserve University [4, 5]. Prof. Thomas Milner from the University of Texas at Austin is currently building a new ODT research group. His group seeks to run ODT systems at much higher image (frame) rates than already existing systems.

The short-time Fourier transform (STFT) is used in the previous generation of ODT systems to extract the information from the recorded IFP's. It is a good choice for the time-frequency representation of ODT signals because of its high time-frequency resolution of modulated signals [6]. The grayscale value of the pixel at the location (i, j) in ODT velocity images is proportional to the flow direction vector at that point through the Doppler shift. The constant of proportionality is a function of the wavelength of the source and the angle between the source wave vector and the flow direction vector. The Doppler shifts are calculated as the difference between f_c and the centroid of the STFT of the IFP obtained by imaging point (i, j).

III. Motivation for a New Algorithm

While the previous generation ODT systems took several minutes to produce one frame, Prof. Milner's group is working on a system, which will capture 10 frames/second, where each frame is 100×100 pixels. With a scan efficiency of 25%, this translates into 2.5 μs/pixel of IFP recording time. Since Doppler shifts in ODT typically range from 0.1 to 3 kHz, each IFP will contain only a small fraction of the period of the Doppler shift frequency (~1μs $\Leftrightarrow$ 0.0013 periods at f_d = 500 Hz). The short time length of the records and the noise of the recorded IFP effectively limit the resolution of STFT-based algorithms.

Project requirements are to develop a new algorithm that could detect flow in these new fast (5-10 frames/second) ODT system in real-time. This algorithm needs to be able to detect 0.1-3 kHz Doppler shifts in a 1 MHz carrier wave using data records of only 1 μs in a noisy environment.

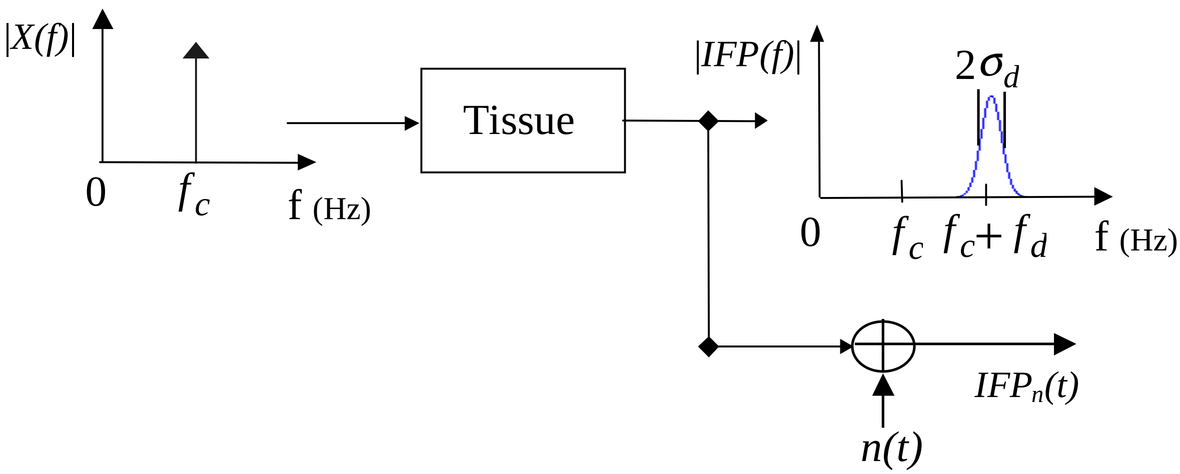

Figure 1: Proposed model for ODT systems

IV. Theory

The proposed model of the system is given in Figure 1 where *X(f)* is the Fourier transform of the source signal, $IFP_n(t)$ is the recorded IFP, $n(t)$ is independent, identically distributed Gaussian noise with zero mean and σ^2 variance, f_c+f_d is the mean of the Doppler spread of variance σ_d^2, and f_c is the frequency of the source signal. The model of $IFP_n(t)$ driven by the physics of the problem is given in (1)

$$IFP_n(t) = \sum_{i=0}^{\infty} W_i \cos(2\pi f_c t + 2\pi f_i t + \phi_i) + n(t) \qquad (1)$$

where ϕ_i is uniformly distributed from 0 to 2π, W_i are Gaussian pulse samples, and f_i is a frequency found in the spectrum of the backscattered light. It can be shown that the behavior of this system can be modeled using an AM-FM model given by (2)

$$IFP_n(t) \cong A(t)\cos\left[2\pi\left(f_c + f_d\right)t + \phi(t)\right] + n(t) \tag{2}$$

Models given by (1) and (2) can be tied together using the constraint given by (3)

$$A(t)\exp\left[j\phi(t)\right] = 2w(t) * \Im^{-1}\left\{\exp\left[j\xi(f)\right]\right\} \tag{3}$$

where $\Im\{x\}$ denotes the Fourier transform operator on $x, w(t) = \Im^{-1}\{W(f)\}$ with $W_i = W(f)\big|_{f=i\Delta f}$ and with $\phi_i = \xi(f)\big|_{f=i\Delta f}$ where Δf is the spacing between the successive samples in the frequency domain. Both $A(t)$ and $\phi(t)$ should be slowly varying functions compared to $\cos(2\pi f_c t)$.

V. Proposed Algorithm

From the physics of the problem, it is known that $\phi(t) = \phi(t-\Delta t)$ for some small Δt where $\Delta t \approx \frac{1}{\text{velocity of blood}} < 30\,ms$ (Δt=1 ms in this paper.) Thus, this unknown phase is eliminated in the cross-correlation of $IFP_n(t)$ and $IFP_n(t-\Delta t)$. The cross-correlation will have the form given in (4)

$$R_T(\tau, \Delta t) \approx \cos\left[2\pi\left(f_c + f_d\right)(\tau - \Delta t)\right] \tag{4}$$

In time Δt, the Doppler shift f_d will exhibit itself as a phase shift, hence the maximum of the cross-correlation will give us the time difference, and thus the phase difference, between the two IFP's. If we assume only one cycle of $R_T(\tau,\Delta t)$, the estimate of f_d is given in (5)

$$\tilde{f}_d = \mathrm{mod}(f_d \Delta t, 1) = \frac{\mathrm{mod}(f_c \Delta t, 1) - f_c \tau_{\max}}{\tau_{\max} - \Delta t} \tag{5}$$

where $\tau_{\max} = \arg\max_{\tau} R_T(\tau, \Delta t)$. Finding this estimate for every pixel produces an ODT velocity image of the tissue.

The histogram of the estimates of Doppler shift for 10,000 trials is shown in Figure 2. The histogram has roughly Gaussian shape with the mean of 300 Hz, which was the Doppler shift of the data in all the trials.

Equation (5) produces a value for a single pixel. The magnitude of the flow direction vector will change smoothly across the image due to the mechanics of uncompressible fluid flow. Thus, we can use the spatial correlation between pixels to refine the estimate of the flow vector. The spatial correlation will eliminate spurious peaks resulting from estimations as given in Figure 2. Taking advantage of the high spatial correlation in ODT velocity images greatly improves the estimates of the flow direction vectors.

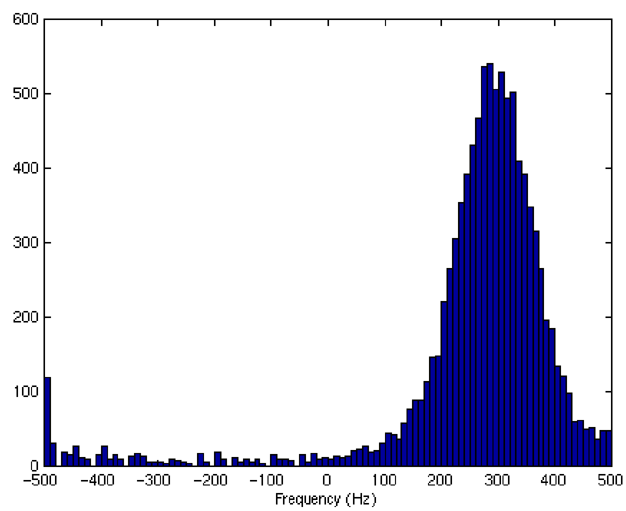

Figure 2: Histogram of $\tilde{f}_d$ at one pixel for 10,000 trials

VI. Performance

The range of detectable Doppler frequencies for a single point is easily obtained to be

$$-0.5 < f_d \Delta t \le 0.5 \qquad (6)$$

This limited frequency range causes the wrap-around effect seen in Figure 2 around –500 Hz. This range in (6) can be greatly expanded by using one-dimensional phase unwrapping technique with the limitation that the gradient along any column of the image cannot exceed $\pm\frac{1}{2\Delta t}$. This constraint can be relaxed by using a two-dimensional phase

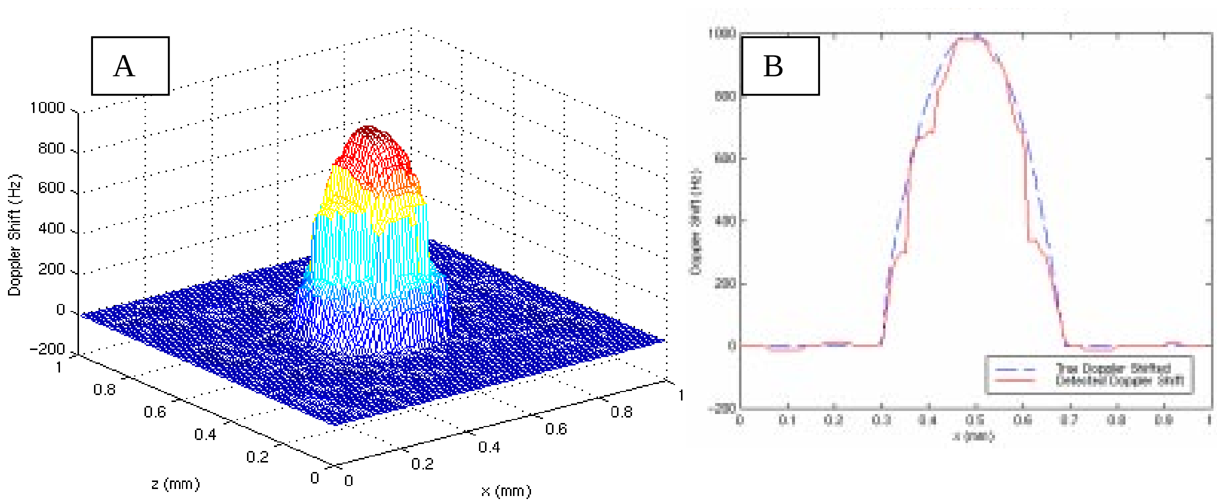

Figure 3: A. 3-D Reconstruction of the blood flow, B. Axial Profile of the Estimate

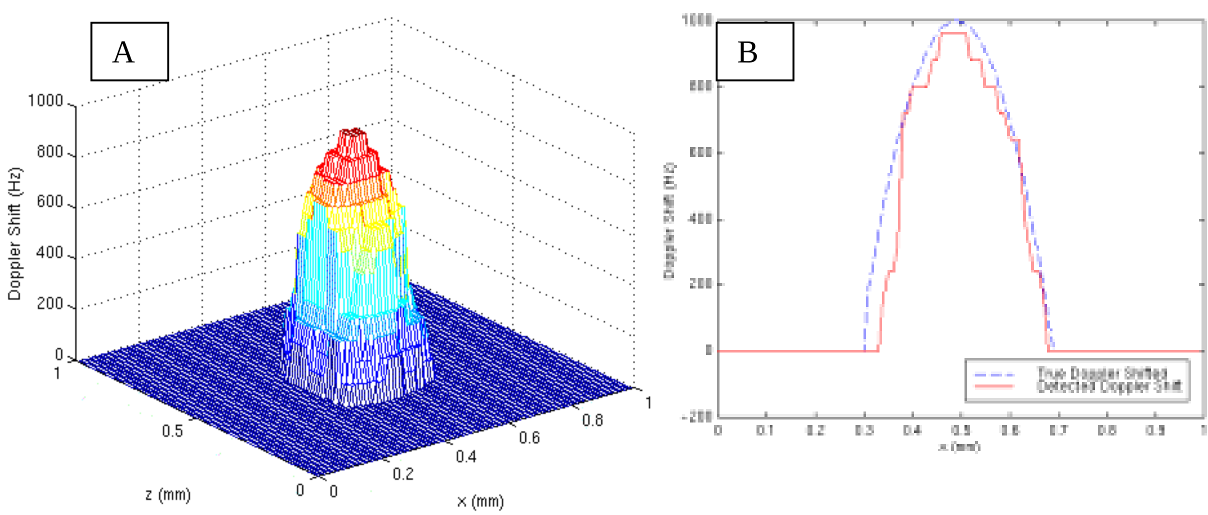

Figure 4: A. 3-D Reconstruction of the blood flow, B. Axial Profile of the Estimate

unwrapping technique that uses gradient in both dimensions. A great body of work has been done in phase unwrapping starting with the landmark paper [7].

The frequency resolution Δf_d of the estimate is derived to be $\Delta f_d = \frac{f_c}{f_s \Delta t}$ where f_s is the sampling frequency. The frequency resolution of the algorithm depends on parameters that we control and allows latitude in the choice of those parameters. The computational cost of the algorithm is mainly dominated by the N_s^2 multiplications in the cross-correlation and the N_s comparisons needed to find the maximum of the cross-correlation, where N_s is the number of samples.

The algorithm gives accurate results even with significant noise, because noise has little effect on the cross-correlation step of the algorithm. The results of the following section are obtained using SNR=-3 dB.

VII. Simulation Results

The following results have been obtained from IFP data simulated to represent a blood vessel of 200 μm radius with its center in the center of the image. The size of images is 100×100 pixels, each pixel 10×10 μm. In the simulated data, the carrier frequency is 1 MHz and Δt is 1 ms. The data was simulated and processed in MATLAB 5.0. The dashed line in Figures 3 and 4 shows the ideal curve of the blood flow. The ideal curve follows the well-known parabolic fluid flow equation. All of the results are obtained using data simulated via the physical model (1) and reconstructed via AM-FM model (2). The reconstructed vessel in Figure 3 used 32 bits precision with 100 MHz sampling rate. At this sampling rate, recording of circa 1 μs (the period of f_c) required 128 samples/pixel. The same vessel is simulated using 4-bit precision and a 12.5 MHz sampling rate (16 samples/pixel.) The results of the reconstruction from this data set are shown in Figure 4.

VIII. Conclusions

We have developed an algorithm that is derived from an AM-FM model of the interference fringe patterns in optical Doppler tomography. The proposed algorithm satisfies and exceeds the challenges put forward by the project requirements. It is not only able to detect the flow, which was the first and foremost requirement of the project, but it is also able to quantify the flow direction vector with notable accuracy even in noisy environment with very short data records. The algorithm is computationally efficient and has potential to be implemented real-time in the new generation of optical Doppler tomography systems with high data acquisition rates.

IX. References

1. B. S. Rinkevichyus and V. I. Smirnov, "Optical Doppler method for investigating turbulent flow by spectral analysis of a signal," *Sov. J. Quantum Electron.*, vol. 3, no. 2, pp. 146-148, 1973.
2. X. J. Wang, T. E. Milner, and J. S. Nelson, "Characterization of fluid flow velocity by optical Doppler tomography," *Optics Letters*, vol. 20, no. 11, pp. 1337-1339, 1995.
3. Z. Chen, T. E. Milner, D. Dave, and J. S. Nelson, "Optical Doppler tomographic imaging of fluid flow velocity in highly scattering media," *Optics Letters*, vol. 22, no. 1, pp. 64-66, 1997.
4. K. D. Manish, T. G. Van Leeuwen, S. Yazdanfar and J. A. Izatt, "Velocity accuracy enhancement and frame rate limitations in color Doppler optical coherence tomography," *Proc. IEEE Conf. Lasers and Electro-Optics*, San Francisco, CA, May 1998, pp 125-127.
5. J. A. Izatt, K. D. Manish, H. Wang, K. Kobayashi and M. V. Sivak, "Optical coherence tomography and microscopy in gastrointestinal tissues," *IEEE J. Selected Topics in Quantum Electronics*, vol. 2, no. 4, pp. 1017-1028, 1996.
6. D. L. Jones and T. W. Parks, "A resolution comparison of several time-frequency representations," *IEEE Trans. Sig. Proc.*, vol. 40, no. 2, pp. 413-420, 1992.
7. J. M. Tribolet, "A new phase unwrapping algorithm," *IEEE Trans. on Acoustics, Speech, and Signal Processing*, vol. 25, no. 2, pp. 170-177, 1977.