

Frequency Domain Optical Tomography Instrument with High Frequencies for Imaging Small Geometries

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Abstract: We present a frequency domain optical tomography instrument for imaging small geometries. The instrument employs modulation frequencies up to 1 GHz which allows yields better separation of absorption and scattering and more accurate reconstructions.

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1. Introduction

Optical Tomography is an emerging imaging modality [1]. It uses non-ionizing light in the visible to infrared range to non-invasively measure the spatial distributions of the optical absorption coefficient (μ_a) and scattering coefficient (μ_s). Researchers are increasingly using small animals as models for diseases and are using optical tomography as a tool to image these animals and explore physiological changes in vivo during disease progressing and over the course of various treatment regimens [2]. With optical tomography, tissue is illuminated and the light that exits the tissue at different positions is measured in order to determine how much light has been attenuated by the optical absorption and scattering inside tissue. Mathematical models of light propagation in tissue are used to reconstruct the tissue's spatial distributions of absorption and scattering from the collected measurements [3].

A common problem encountered in DOT is that the image reconstruction is ill-posed and ill-determined which can have a negative impact on the spatial resolution and accuracy of the reconstructed images. One method of reducing the ill-posedness, is to acquire DOT data in either the time domain or frequency domain instead of the steady-state-domain. We therefore consider here a frequency-domain system. Previously it has been shown that modulation frequencies greater than 400 MHz are often optimal [4]. In the following section we present the design of a frequency domain optical tomography system that is capable of using high modulation frequencies up to 1 GHz, which allow for better separation of absorption and scattering and more accurate reconstructions. Details on signal modulation and demodulation, as well as data calibration are described.

2. Frequency domain optical tomography system

2.1 System description

Figures and tables should be centered (except for small figures less than 2.6 in. or 6.6 cm in width, which may be placed side by side) and located inside paper margins. Figure 1 shows a diagram of the frequency domain instrument setup in our laboratory. The system employs two laser diodes at 2 different wavelengths $\lambda_1 = 757$ nm and $\lambda_2 = 828$ nm. The laser diodes are pigtailed into single mode optical fiber. They both are driven by modulated laser drivers and have peltier cooling. The 2 laser diodes have 3 dB bandwidths of 800 MHz and 700 MHz respectively and 10 dB bandwidths of 1.3GHz. Those 2 fibers are then input to a 2x32 fiber optic switch (that has low coupling losses and fast switching times (4 ms)). The fiber optic switch uses standard 62.5 μ m graded-index multimode fibers which provide good coupling efficiency from the single mode fibers and high bandwidth. Both source and detector fibers interface with the imaging head in which the animal is placed. Detector fibers that deliver light from the imaging head have their tips arranged hole drilled into a threaded aluminum camera lens cap. The cap is attached to a 50 mm T-mount extender tube that is attached to the camera lens in order to fix the ends of the fiber near the minimum focusing distance of the lens. A "close-up" lens (60mm f/2.8D) is used to maximize the amount of light that is focused into the intensifier tube of our intensified CCD camera system (Picostar HR LaVision, Germany).

The intensifier has 3 stages. First light passes through the input window and is incident on the photocathode. The photocathode, when activated by a voltage, converts photons into electrons. These electrons enter and accelerate through a micro channel plate (MCP). The MCP produces a gain, which depends on the voltage which is applied across the MCP. The electrons exit the MCP and are incident on a phosphor screen where they are converted back to photons. The photons exiting intensifier are then focused on to the CCD camera. The CCD camera has 1376 x 1040 pixels and is triggered by and transfers the acquired images to a PC running DaVis imaging software (LaVision, Germany). Since the detector fibers are only focused onto a small area, the CCD crops the data to a smaller rectangle showing the detector fibers. For each fiber, the pixels showing light from that fiber are averaged

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to produce one detector value for that fiber. Conventionally a photocathode is gated by a pulse so the intensifier can act like a shutter. In our instrument a sinusoidal voltage is applied across the cathode allowing RF modulation. Similar instruments are described in [4], but they operate at lower frequencies and have longer acquisition times. The photocathode voltage signal, MCP voltage and other bias voltages are supplied by the High Rate Imager.

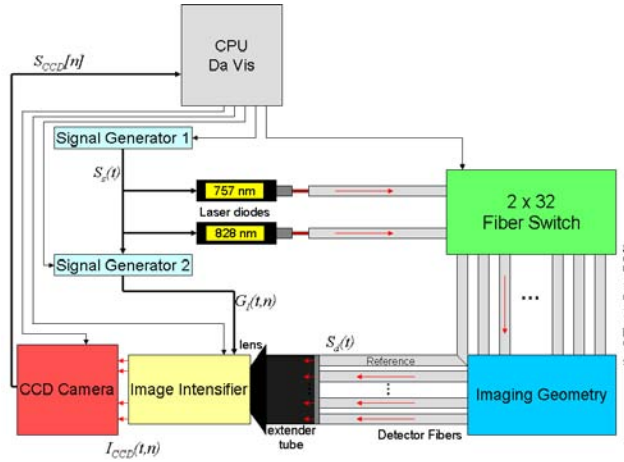


Fig. 1: Diagram of the frequency domain optical tomography instrument

2.2 Signal details

Two signal source generators are used to produce the RF signals that are needed for modulating the laser diodes and the photocathode of the image intensifier. The parameters of the generated signals (amplitude frequency, phase etc.) are computer controlled through the serial port. One signal generator is master and its RF signal is input to the both of the modulated laser drivers. The frequency of the slave signal generator is set to that of the master signal generator. The phase difference between the resulting 2 signals is controlled, which as we will show is necessary to demodulate the optical signal. The current driving laser diodes can be electronically modulated in order to produce modulated optical source. Since photon density can only be positive, sinusoidal modulation of the optical source signal must take the form of a raised sinusoid: $S_s(t) = A_s(1 + m_s)\cos(\omega t + \phi_s)$ (1).

The raised sinusoid is always positive and thus, m_s is less than 1. As the photon density waves travel in tissue, the optical properties will have a system response $H_t(\omega)$, which will consist of a frequency dependant attenuation and phase shift. The measured detector signal will thus take the form of:

$$S_d(t) = A_s(H_t(0) + |H_t(\omega)|m_s \cos(\omega t + \phi_s + \phi_t)) \quad (2)$$

We modulate the gain of an image intensifier in order to partially demodulate the signal in the optical domain. The gain of the intensifier G_I , is also a raised sinusoid which prevents standard homodyne demodulation techniques, but we take several, n , measurements and while modulating the gain, incremental phase shifts, $\Delta\phi_I$, are added.

$$G_I(t, n) = A_I(1 + m_I \cos(\omega t + n\Delta\phi_I)) \quad (3)$$

The light intensity that is incident on the CCD, I_{CCD} , is a product of the detector signal and the gain signal, and contains 2 DC terms and many high frequency terms (HFT). The light incident on the CCD is integrated over the exposure time. Since the exposure time is much longer than the modulation period, this leaves only the DC terms.

$$S_{CCD}[n] = \int_{t_e} I_{CCD}(t, n) dt = A_I A_s t_e H_t(0) + A_I A_s t_e \frac{m_I m_s}{2} |H_t(\omega)| \cos[\phi_s + \phi_t - n\Delta\phi_I] \quad (4)$$

Because the product of the S_d and G_I is sampled at discrete phase offsets this technique can be thought of as a phase sampling heterodyne technique that produces a discrete signal with N_s samples. It has however previously been referred to as homodyne detection [5]. The DC and AC responses can be found by applying the appropriate digital filter to (4).

$$M_{DC} = A_I A_s t_e H_t(0) = \sum_{n=1}^{N_s} \frac{1}{N_s} \times S_{CCD}[n] \quad (5)$$

$$M_{AC} = A_I A_s t_e \frac{m_I m_s}{2} |H_t(\omega)| = \left| \sum_{n=1}^{N_s} \frac{2}{N_s} (\cos[(n-1)\Delta\phi_I] + j \sin[(n-1)\Delta\phi_I]) \times S_{CCD}[n] \right| \quad (6)$$

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$$M_{\phi} = \phi_s + \phi_t = \arg \left(\sum_{n=1}^{N_s} \frac{2}{N_s} (\cos[(n-1)\Delta\phi_I] + j \sin[(n-1)\Delta\phi_I]) \times S_{CCD}[n] \right) \quad (7)$$

Absolute measurement can be obtained by taking measurements of a reference medium with a known response $H_r(\omega)$ before imaging a target in order to account for unknown coupling coefficients, $A_t, A_s, t_e, m_1, m_2, \phi_s$. This is evident because by the following equations where the unknown coupling coefficients can be eliminated leaving the system responses of interest.

$$H_t(0) = H_r(0) \times M_{DC}^T \quad M_{DC}^R \quad (8)$$

$$|H_t(\omega)| = |H_r(\omega)| \times M_{AC}^T \quad M_{AC}^R \quad (9)$$

$$\phi_t(\omega) = \phi_r(\omega) + M_{\phi}^T - M_{\phi}^R \quad (10)$$

3. Results

An initial phantom experiment was conducted with a 3.2 cm diameter cylinder consisting of 1% Intralipid and a rod inclusion consisting of higher absorption. Data was collected using 25 sources and 25 detectors located around the cylinder along a 1.6 cm height. Measurements were taken at DC and a modulation frequency of 600 MHz. Fig. 2 show phantom (a), and initial 3D reconstruction results (b) with a diffusion code. Initial results are promising and are expected to improve as the system and reconstruction routine are optimized.

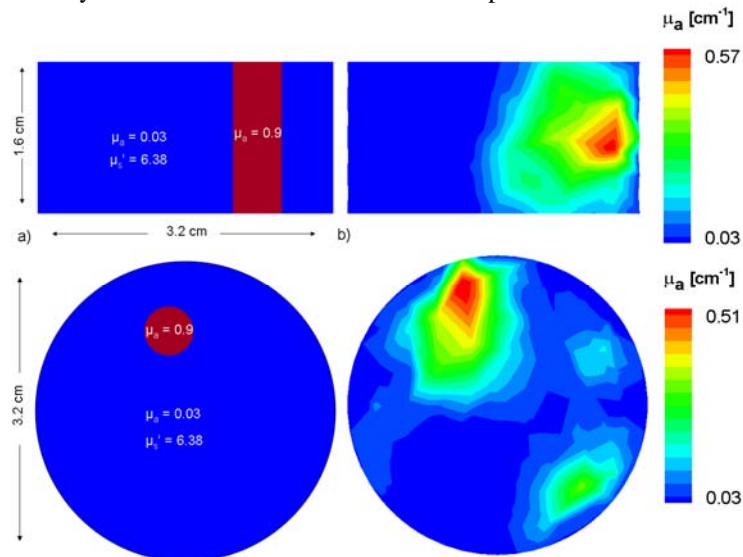


Fig 2: Actual (a) and Reconstructed (b) images for the phantom experiment

4. Conclusion

In this work we have described a frequency-domain instrument that uses high modulation frequencies for imaging small animals. Unlike existing systems, our instrument is capable of source-modulation frequencies up to 1 GHz. We found that initial experiments with the system yield promising reconstructions. This work was supported in part by the National Institute of Biomedical Imaging and Bioengineering (NIBIB grant 5R01-EB001900, Hielscher) which is part of the National Institute of Health.

5. References

- [1] A. P. Gibson, J. C. Hebden, S. R. Arridge, "Recent advances in diffuse optical imaging," *Phys. Med. Biol.* **50** pp. R1-R43 (2005).
- [2] A.H. Hielscher, "Optical tomographic imaging of small animals," *Current Opinion in Biotechnology* **16**(1), pp. 79-88 (2005).
- [3] S. R. Arridge, "Optical tomography in medical imaging", *Inverse Problems* **15** pp. R41-R93 (1999).
- [4] X. Gu, K. Ren, A. H. Hielscher, "Frequency-domain sensitivity analysis for small imaging domains using the equation of radiative transfer," *Appl. Opt.* **46**, 1624-1632 (2007).
- [5] J. S. Reynolds, T. L. Troy, E. M. Sevick-Muraca, "Multipixel Techniques for Frequency-Domain Photon Migration Imaging," *Biotechnol. Prog.* **13** pp. 669-680 (1997).