Theoretical Study of Combined Acousto-optical Tomography and Slow Light Filters

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Abstract: Monte Carlo simulations were used to determine the contrast-to-noise ratio of acousto-optical tomographic imaging with slow light filters versus possible imaging depth. Both reflection and transmission setups were considered. The theoretical model showed that imaging through 12 cm of breast tissue could be plausible.

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1. Introduction

Optical imaging of biological tissue has good molecular contrast but poor spatial resolution and penetration depth of near infrared light. Acousto-optical tomography (AOT) has the potential to overcome these current imaging limitations. AOT illuminates light through an ultrasound (US) focus in a medium, where photons become “tagged” [1-3]. The tagged light can then be detected using various methods, such as speckle contrast analysis, heterodyne digital holography, photorefractive crystal interferometry, and spectral hole burning methods [1, 2].

We recently published a review of AOT methods [1]. The field has begun roughly the same time as a photoacoustic tomographic (PAT) imaging, but has not gained the same traction and has remained mostly in phantom imaging stage. We identified several challenges that the field still faces to push AOT into in vivo imaging. For example, the decorrelation time of tissue is <1 ms and imaging must take place within this time limit. Also, there needs to be imaging within the tissue optical window, most studies in the review used between 532-638 nm or >1000 nm lasers for imaging [1]. However, despite these challenges, we believe that AOT holds the potential for deep tissue optical imaging. One group used photorefractive crystals and two-wave mixing to image through a 9.4 cm tissue phantom [4]. Another study imaged through a 9 cm tissue phantom, by using slow light filters (SLFs) [5]. Continuing with the concept of combined AOT and SLFs, a theoretical study showed that the imaging depth of AOT could be twice that of PAT in a reflection setup [3]. The study was expanded to use Monte Carlo (MC) simulations instead of the 1D diffusion approximation with boundary conditions and looked at both reflection and transmission geometries. The theoretical study found that imaging through a ~12 cm slab of tissue may be possible with SLFs [6].

SLFs are created by using rare earth crystals and spectral hole burning methods. By pumping the rare earth crystal with a particular laser frequency (i.e. the tagged light frequency) the crystal becomes “transparent” to that frequency of light and absorbs other frequencies. By combining slow light effects, the tagged light also travels slower through the crystal (~μs slower) and aids in the separating the tagged and untagged light [5]. In the previously mentioned SLF study, the untagged light suppression was 30 dB [5]. However, advancements have been made and now the SLFs have a background light suppression of 60 dB, but still have not be utilized in AOT applications yet [7].

We have continued to investigate theoretically the ability to combine spectral hole burning (SHB) into an AOT systems using MC simulations. The MC simulations were used to estimate how many photons would travel through a US focus and to a detector. The number of photons are compared with the number of photons from a typical reflection measurement. Therefore, we calculated the contrast-to-noise ratio (CNR) for both reflectance and transmittance configurations to better understand how much contrast can be achieved deep within biological medium. For this study, we concentrate on simulations that would imitate breast imaging.

2. Methods

Previously, we looked at imaging through muscle to obtain measurements on the heart [6]. Here, we examine breast tissue and the ability to distinguish between a localized increase in absorption coefficient (50% increase from background μ_a) and the background absorption. The source was assumed to be an 880 nm pulsed laser. There were two possible experimental setups taken into consideration: reflectance and transmittance (see Fig. 1). In the reflection setup the laser, US transducer, and detector are on the same side of the medium and the US focus is translated within the medium. The US transducer is placed halfway between the laser and detector. For the transmittance setup, the laser and detector are on opposite sides of a medium of a given length (L). The US transducer is moved laterally and the US focus is moved z_{US} distance from the laser source.

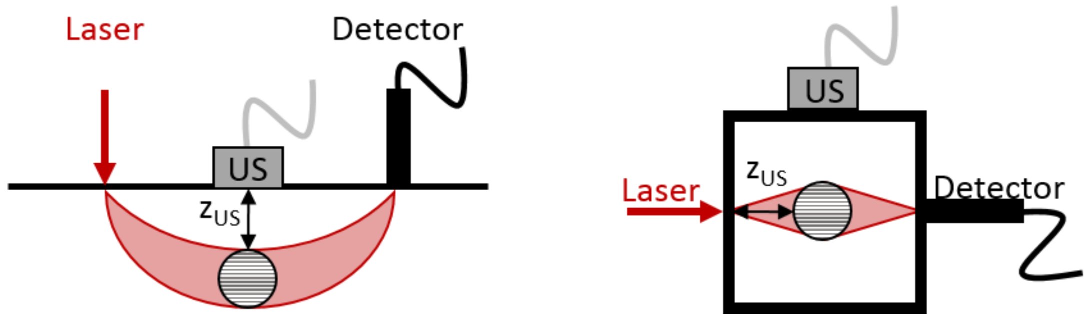

Figure 1 Reflection (Left) and transmission (Right) setups for AOT/SLF simulations. The circle represents the ultrasound focus. US: ultrasound

Monte Carlo simulations were run using parallel computational code that uses the GPU of the computer to effectively decrease computation time [8]. The simulation parameters can be seen in Table 1. The theoretical model is described in reference [6]. Briefly, the fluence map, diffuse reflectance, and diffuse transmittance measurements from the MC simulations are used to determine the number of photons that travel from the source to detector, but also pass through the ultrasound focus (P_{sig}). To obtain the CNR, two signals were calculated: with and without an increase in absorption of the background medium, P_{sig1} and P_{sig2}, respectively. The CNR was calculated using:

$$CNR = \frac{|P_{sig2} - P_{sig1}|}{\sqrt{P_{sig1} + P_{bkg}}} \tag{1}$$

where P_{bkg} are all the untagged photons that were detected.

Table 1 Monte Carlo simulation parameters

Monte Carlo Parameters	Breast@880 nm[9]
Photon packages in MC	10^{12}
Absorption coefficient	0.08 cm^{-1}
Scattering coefficient	100 cm^{-1}
Transport scattering coefficient	10 cm^{-1}
Anisotropic factor	0.9
Index of refraction	1.37

3. Results

The CNR for the reflectance setup (Fig. 2 Left) decreases as imaging depth increases. The CNR remains above 1 for to about 6 cm when the source-detector (SD) distance was 2 cm and 4 cm. As for the simulation with SD distance of 6 cm, CNR remains above 1 until about 5.7 cm.

For transmittance, as the length of the medium increased, the CNR decreased. Furthermore, the upper limit for the 10 cm slab had a CNR>10 at every imaging depth. Last, there was a CNR >1 for all depths of the 12 cm slab medium. This simulations showed that 12 cm imaging may be possible if sufficient absorption contrast is attained.

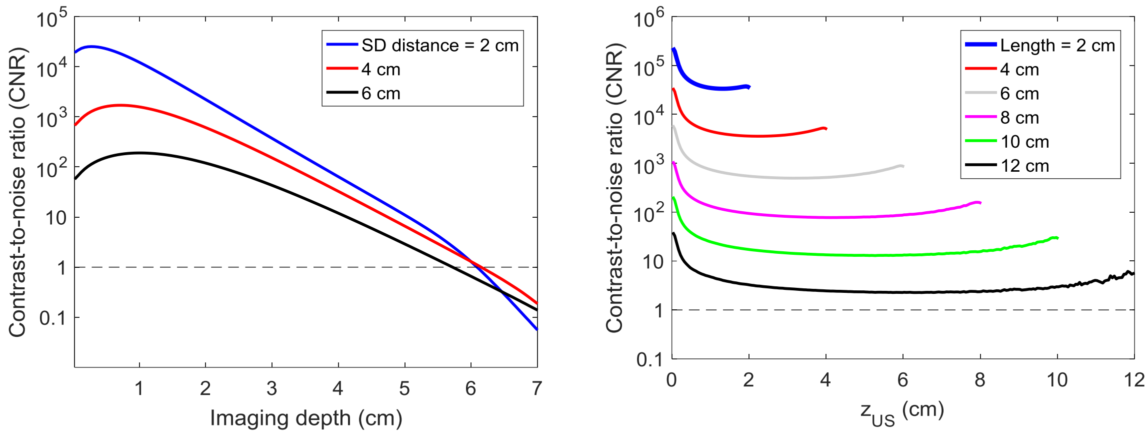

Figure 2 The CNR for reflection (Left) and transmission (Right) setups. The reflection setup shows the CNR for three different source-detector (SD) distances in which the imaging depth is the location of the ultrasound focus in the medium. The transmission setup shows the CNR for different depths of the ultrasound focus (z_{US}) at different lengths of the medium (i.e. 2 cm, 4 cm, etc.). The dash line represents where CNR = 1.

4. Discussion

The general trends of our results are very similar to what was previously observed [3, 6]. However, AOT has shown to image at deeper depths than through muscle [6]. In the current study, a maximum imaging depth with a CNR=1 for reflection geometry was 6 cm. Even with the shorter source-detector distances, the deeper imaging depths were still able to have sufficient contrast to distinguish the two absorptions. The smaller SD distances were more efficient near the boundary of the medium but as the imaging depth increase there was less of a difference in CNR among the different SD distances. Yet, AOT is specific in the sense that if the photon is tagged, then one of the locations of the photon within the medium is known. More investigation is necessary to understand optimal SD distances for reflection AOT.

Also, a slab of length 12 cm would still have sufficient contrast (CNR>1). The CNR remained relatively constant (except near the boundaries) as the ultrasound focus depth was changed, which is unlike the reflection setup. For the reflection setup light not only travels into the medium but then back out again in which case the light path would be a bit more than double the imaging depth. While in the transmission setup, the light travels through the medium [6].

These are promising results that show the possibility of deep tissue imaging when combining spectral hole burning methods with AOT. Yet, phantom studies are still necessary to validate these results.

5. Acknowledgments

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