# Near Infrared Optical Tomography at MR resolution: Photo-magnetic Imaging

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***Purpose:***

MR thermometry is an advanced non-invasive technology with many clinical applications that can accurately measure temperature tomography in vivo [1]. Here, a combination of MR thermometry and diffuse optical tomography (DOT) has been proposed to develop a new multi-modality system, photo-magnetic imaging (PMI). DOT has an advantage in that it can monitor different chromophore concentrations such as hemoglobin and de-oxygen hemoglobin to provide functional information and shed light on some of the biomarkers of cancer like angiogenesis and hypermetabolism [2,3]. However, the high scattering nature of near infrared light in biological tissue contributes to the ill-posed and undetermined nature of the inverse problem resulting in poor spatial resolution and low quantitative accuracy. Optical detectors can only measure the photon flux at the boundary of the subject which makes it difficult to reconstruct three dimensional volume optical properties from two dimensional measurements. Instead of using conventional optical sensors, PMI uses MRI as the detector to provide 3D measurements. In fact, with just the addition of a continuous wave laser, PMI can be a simple addition to any existing MR system and enhance cancer diagnosis and treatment by providing MR resolution optics tomography.

***Method:***

In the conventional DOT system, photons from external near infrared laser illuminate the tissue, propagate through the medium, and are detected by photon sensors at the surface. Instead of using conventional optical sensors to measure the photon flux density around the subject boundary, PMI uses MRI to measure the temperature change which is directly proportional to the amount of photons absorbed by the tissue due to the energy conversion of light to heat. A finite-element based photon diffusion model and bio-heat diffusion equation is used to model the photon distribution and temperature change in the tissue. Solving the inverse problem with the Levenberg-Marquardt algorithm, a high resolution optical absorption tomography map of the subject can be reconstructed. The experiment setup is presented in Figure 1. A 808nm 30W laser controlled by Winvue-HCT software with temperature feedback control in the MR operation room was connected to a 15 m long, 1 mm diameter optical fiber. The other end of the fiber was connected to the top of an animal MRI RF coil to illuminate the subject. A GRIN lens was used at the tip of the fiber to collimate the laser output before irradiating the subject. An agar phantom was used to mimic small mice muscle with an optical absorption coefficient of 0.013 mm-1 using optical dye and placed in the center of the coil. A gradient echo sequence was used for proton resonance frequency shift phase map MR thermometry with a TR and TE of 60 ms and 16 ms, respectively. A consecutive 6 frames dynamic sequence was used to monitor the laser induced temperature change of the agar phantom with the first four frames averaged as the temperature baseline image. The laser was turned on at the 3$^{th}$ frame and turn off at the end of the 4$^{th}$ frame. Consequently, a temperature increase is seen in the 3$^{th}$ and 4$^{th}$ frames due to the absorption of the photons from the laser and decreases in the 5$^{th}$ and 6$^{th}$ frames due to heat diffusion.

***Results and Conclusion:***

Simulated FEM based temperature changes in the tomographic image of the agar phantom was compared to the experiment results at each time point. A 5 mm diameter, with 4x higher absorption than background tissue was embedded in a 40 mm diameter agar phantom. These inclusions act as a vascularized tumor which has a higher optical absorbance compared to normal tissue due to the increased amount of hemoglobin. This leads to a higher number of photons absorbed resulting in an increase in the temperature of the high absorption regions. The results from Figure 2 show that the forward simulation accurately estimates the laser induced temperature change during both heating (laser on) and cooling phase (laser off). In the future, with the addition of the inverse solver, we will be able to reconstruct an optical property map with MR resolution. Additionally, optical contrast agents such as special coating gold nanoparticle or non-targeted optical contrast agent such as Indocyanine Green (ICG) can be monitored in real time by dynamic PMI [4]. By further expanding the prototype PMI system to a multiple wavelength system, tissue chromophore concentration images can be acquired that will make PMI a true functional imaging technique.

***Reference:***

[1] [Rieke, 2008, MR thermometry], [2] [Tromberg, 2008, Assessing the future of diffuse optical imaging technologies for breast cancer management], [3] [Gibson, 2005, Recent advances in diffuse optical imaging], [4] [Holbrook, 2010, Real-time MR thermometry for monitoring HIFU ablations of the liver]

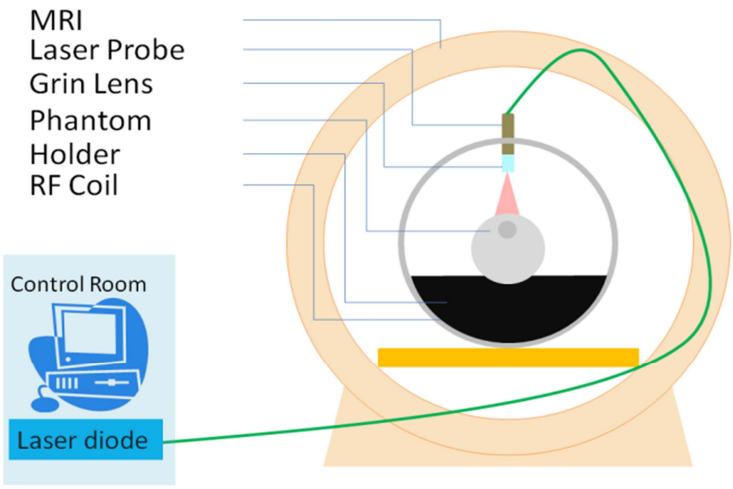


Figure 1. Photo-Magnetic Imaging system.

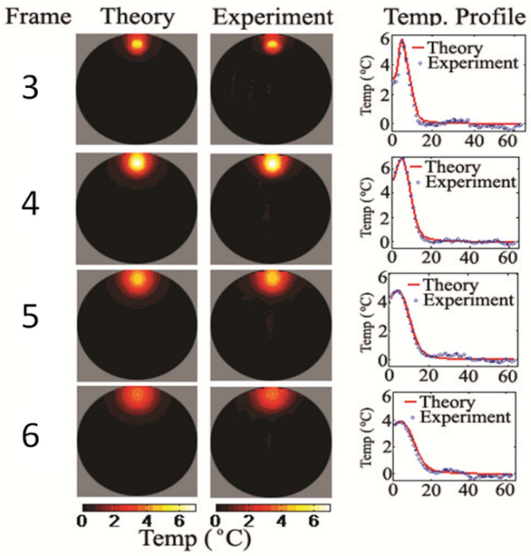


Figure 2. Finite element based simulation (column 1) results for laser induced temperature change of the agar phantom at 4 different time points comparing with the temperature change monitored with MR thermometry. The second column compares the modeling results with the temperature variation maps monitored with MR thermometry. A 40 mm diameter phantom is embedded with a 5 mm diameter inclusion (4x absorption contrast) to mimic a tumor in the biological tissue. The third column shows the phantom temperature profile across the high absorbing inclusions.