Magnetoacoustic imaging of human liver tumor with magnetic induction

Gang Hu,1 Erik Cressman,2 and Bin He1,a
1Department of Biomedical Engineering, University of Minnesota, Minnesota 55455, USA
2Department of Radiology, University of Minnesota, Minnesota 55455, USA

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Magnetoacoustic tomography with magnetic induction (MAT-MI) is an imaging technique under development to achieve imaging of electrical impedance contrast in biological tissues with spatial resolution close to ultrasound imaging. However, previously reported MAT-MI experimental results are obtained either from low salinity gel phantoms, or from normal animal tissue samples. In this study, we report the experimental study on the performance of the MAT-MI imaging method for imaging in vitro human liver tumor tissue. The present promising experimental results suggest the feasibility of MAT-MI to image electrical impedance contrast between the cancerous tissue and its surrounding normal tissues. © 2011 American Institute of Physics. [doi:10.1063/1.3543630]

Electrical impedance imaging approaches have been widely explored for several decades, since changes in tissue’s electrical impedance provide useful physiological and pathological information about the biological tissues. Various imaging techniques have been employed to discern these conditions, including electrical impedance tomography,1,2 magnetic resonance electrical impedance tomography,3,4 magnetic induction tomography,5 magnetooacoustic tomography,6,7 Hall effect imaging,8 and the recently proposed magnetoacoustic tomography with magnetic induction (MAT-MI).9–12 Among these methods, MAT-MI features excellent spatial resolution (near to ultrasound imaging) as well as noncontact magnetic energy deposition. In MAT-MI, the sample is placed in a static magnetic field. Then, a time-varying magnetic field is applied to induce eddy currents in the conductive tissue volume. The Lorentz force is then generated, from the interaction between the eddy current and magnetic field, and causes mechanical vibration of the tissue. Acoustic waves are produced as a result of the vibration, and can be recorded by acoustic probes placed surrounding the object. Previous studies have already demonstrated that MAT-MI is capable and sensitive to imaging electrical conductivity contrast in gel phantoms,10 salted animal tissues,11 and fresh animal tissues12 with a conductivity contrast around 0.4~0.6 S/m. In the present study, we conducted MAT-MI imaging of human liver tumor tissue samples to directly test the feasibility of distinguishing and imaging of cancerous tumor in human by means of MAT-MI.

In MAT-MI, in order to deposit energy, an excitation coil is used to send out a pulsed magnetic stimulation $B_{1}(r,t)$ to a tissue sample (see Fig. 1). Usually, the pulse duration is at microsecond scale for megahertz ultrasound generation. According to Faraday’s law, an electrical field $E(r,t)$ will be induced in the space. If the object is conductive and with conductivity distribution $\sigma(r)$, we have the current density distribution $J(r,t)$ in the object which can be determined by $J(r,t) = \sigma(r)E(r,t)$. The quasistatic condition is satisfied in the MAT-MI problem, as the system working frequency is low. Because of the short duration of the magnetic stimulation, the current density $J(r,t)$ can be approximated as the product of a pure spatial function $J(r)$ and a delta function $\delta(t)$. With a static magnetic field $B_{0}(r)$ and the corresponding Lorentz force $J(r) \times B_{0}(r)$, the acoustic pressure distribution $\rho(r',t)$ at spatial point $r'$ in the acoustically homogenous medium will follow the wave equation.9

![FIG. 1. (Color online) Schematic of MAT-MI for tumor imaging.](image-url)
signals are available, or by using a multi-excitation method. The left hand side of Eq. (1) can be computed using the time reversal method. The right hand side of Eq. (1) is divergence of Lorentz force which represents acoustic source responsible for the acoustic signals recorded by a probe.

For this experiment, we used an experimental system with a circular measurement configuration (Fig. 1). Both the sample and the transducer were immersed under distilled water for acoustic signal coupling. The flat transducer was mounted on a holder driven by a step motor for mechanical rotation in the XY plane. This transducer had a diameter of 25 mm and a peak sensitivity frequency close to 0.5 MHz. For higher detection sensitivity in acoustic measurements, the transducer was custom fabricated using piezocomposite technology. Piezoelectric signals collected by the transducer were amplified with a low-noise ultrasound amplifier with 90 dB voltage gain before the data digitalization. In order to improve signal-to-noise ratio, a customized magnetic stimulator was employed to deliver strong yet safe magnetic energy to the object. The magnetic coil with outer diameter of 70 mm was placed under the water tank. The coil driver was equipped with an adjustable high-voltage source, a capacitor, and a solid state switch for handling large currents. Fast discharge of the internal capacitor produced stimulation with a pulse width of 1.6 μs. When the charging voltage was set at peak voltage (24 kV), the magnetic stimulator was able to induce an average electrical field of 550 V/m at the bottom of the sample. To build a static magnetic field, two permanent magnets with diameter of 75 mm were placed on the top and bottom of the sample. The separation between the two magnets was adjustable according to the sample thickness. The produced magnetic flux density was 0.2–0.3 T. A data acquisition system was used for trigger control, signal synchronization and data transferring. For each channel, 2048 data points were collected with a 5 MHz sampling rate.

All experimental studies have been approved by the Institutional Review Board (IRB) for Human Research at the University of Minnesota. An excised fresh liver sample with original size of $30 \times 40 \times 12$ mm$^3$ was obtained from a patient having previously undergone surgical removal of liver tumor. Pathology study of the liver tumor samples confirmed that the tumor sample was a cancerous tumor. The tumor site in the sample was located approximately at the center, presenting an irregular shape with maximum diameter of roughly 1.5 cm. We built a phantom consisting tissues with regular shape for this experiment. The tumor site was excised from the surrounding normal tissue. Direct conductivity measurements were performed for both tumor tissue and the surrounding tissue in ten different points (five for tumor tissue and five for normal tissue) at 0.5 MHz before and after the excision by using a four-electrode probe. The probe constant was precalibrated with standard solutions (Oakton Inc.) of varying nominal conductivity values of 0.26, 0.83, and 1.19 S/m at an experimental temperature of 21 °C. The calibrated results indicated good measurement linearity for a conductivity range from 0.20 to 1.20 S/m, which was comparable to that of fresh soft tissues. No significant changes in the electrical conductivity had been observed by the tumor excision. The tumor tissue had conductivity of 0.65–0.70 S/m, while the normal tissue characterized a lower conductivity of 0.25–0.28 S/m. Following conductivity measurements, the tumor and normal tissue samples were diced into rectangular shapes with sizes of $11 \times 4.5 \times 8$ and $11 \times 5 \times 8$ mm$^3$, respectively. Lastly, a tumor phantom was composed by pressing the two tissue portions together as shown in Fig. 2(a).

In the imaging experiments, the transducer collected the MAT-MI ultrasound signals with angular step of 2°. At each view point, 200 times data averaging was used. We reconstructed the acoustic source distribution by using the two-dimensional discrete form of Eq. (2). Figure 2(b) shows the reconstructed acoustic source image, which illustrates contours being consistent with the boundaries between the tumor and normal tissue, as well as the tissues with the surrounding gel [Fig. 2(a)]. Some feature points at the tissue interface and outer boundary were marked for the comparison. Both the location and the size of the sample were well distinguished and imaged. For a comparison, we did a control experiment on a phantom, which was made by pushing two pieces of
The contact interface are observed in the reconstructed image the same conductivity value. No significant signals from that shown in Fig. 3 using conductivity differences as a means for contrast. Hence, such vertical line and other boundary lines represent mechanical discontinuities contribute little to MAT-MI signals. Consequently, Fig. 2(a) indicates the feasibility for imaging real human liver cancerous tumor in an in vitro setting by means of the MAT-MI.

In conclusion, we demonstrate that the MAT-MI imaging method is able to distinguish with high spatial resolution the small electrical conductivity contrast formed by liver tumor tissues and normal tissues. Further investigation may establish MAT-MI as an important technique for imaging and earlier detection of cancers in a clinical setting.

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1K. Paulson, W. Lionheart, and M. Pidcock, IEEE Trans. Med. Imaging 12, 681 (1993).
2P. Metherall, D. C. Barber, R. H. Smallwood, and B. H. Brown, Nature (London) 380, 509 (1996).
3M. Joy, G. Scott, and M. Henkelman, Magn. Reson. Imaging 7, 89 (1989).
4O. Kwon, E. Woo, J. Yoon, and J. K. Seo, IEEE Trans. Biomed. Eng. 49, 160 (2002).
5A. J. Peyton, Z. Z. Yu, G. Lyon, S. Al-Zeibak, J. Ferreira, J. Velez, F. Linhares, A. R. Borges, H. L. Xiong, N. H. Saunders, and M. S. Beck, Meas. Sci. Technol. 7, 261 (1996).
6B. C. Towe and M. R. Islam, IEEE Trans. Biomed. Eng. 38, 892 (1988).
7B. J. Roth, P. J. Basser, and J. P. Ir Wikswo, IEEE Trans. Biomed. Eng. 41, 723 (1994).
8H. Wen, J. Shah, and R. S. Balaban, IEEE Trans. Biomed. Eng. 45, 119 (1998).
9Y. Xu and B. He, Phys. Med. Biol. 50, 5175 (2005).
10X. Li, Y. Xu, and B. He, J. Appl. Phys. 99, 066112 (2006).
11R. Xia, X. Li, and B. He, Appl. Phys. Lett. 91, 083903 (2007).
12G. Hu, X. Li, and B. He, Appl. Phys. Lett. 97, 103705 (2010).
13Y. Xu and L. V. Wang, Phys. Rev. Lett. 92, 033902 (2004).
14X. Li, Y. Xu, and B. He, IEEE Trans. Biomed. Eng. 54, 323 (2007).
15X. Li and B. He, IEEE Trans. Med. Imaging 29, 1759 (2010).
16Z. Tsai, J. A. Will, S. Hubbard-Van Stelle, H. Cao, S. Tanggikasolmum, Y. B. Choy, D. Haemmerich, V. R. Vorperian, and J. G. Webster, IEEE Trans. Biomed. Eng. 49, 472 (2002).
17Q. Ma and B. He, IEEE Trans. Biomed. Eng. 55, 813 (2008).

FIG. 3. (Color online) (a) Photograph of a real human liver tumor sample, which retains the naturally formed tissue structure. Several markers were added at feature points at tissue interface and boundary for image comparison. (b) Reconstructed MAT-MI image of the sample shown in (a).