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Magnetooacoustic tomography with magnetic induction (MAT-MI) for imaging electrical conductivity of biological tissue: a tutorial review

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Abstract
Magnetooacoustic tomography with magnetic induction (MAT-MI) is a noninvasive imaging method developed to map electrical conductivity of biological tissue with millimeter level spatial resolution. In MAT-MI, a time-varying magnetic stimulation is applied to induce eddy current inside the conductive tissue sample. In the presence of a static magnetic field, the Lorentz force acting on the induced eddy current drives mechanical vibrations producing detectable ultrasound signals. These ultrasound signals can then be acquired to reconstruct a map related to the sample’s electrical conductivity contrast. This work reviews fundamental ideas of MAT-MI and major techniques developed in recent years. First, the physical mechanisms underlying MAT-MI imaging are described, including the magnetic induction and Lorentz force induced acoustic wave propagation. Second, experimental setups and various imaging strategies for MAT-MI are reviewed and compared, together with the corresponding experimental results. In addition, as a recently developed reverse mode of MAT-MI, magnetoo-acousto-electrical tomography with magnetic induction is briefly reviewed in terms of its theory and experimental studies. Finally, we give our opinions on existing challenges.

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and future directions for MAT-MI research. With all the reported and future technical advancement, MAT-MI has the potential to become an important noninvasive modality for electrical conductivity imaging of biological tissue.

Keywords: magnetoacoustic tomography with magnetic induction, MAT-MI, MAET-MI, Lorentz force, conductivity, bioimpedance, imaging

(Some figures may appear in colour only in the online journal)

1. Introduction

Electrical properties of biological tissue including electrical conductivity $\sigma$ and permittivity $\gamma$ are important biophysical parameters in the study of electrophysiology and electromagnetic therapies such as transcranial direct current stimulation (tDCS) (Malmivuo et al 1995, Sadleir et al 2010). In addition, due to their changes under physiological and pathological conditions, tissue electrical properties may serve as an imaging contrast for possible diagnosis and research use (Geddes and Baker 1967). Previous studies have shown that cancerous breast tumor tissue has significantly different electrical properties than normal breast tissue or benign tumors (Surowiec et al 1988, Jossinet 1996, 1998). Significant electrical conductivity difference has also been found between liver tumors and normal liver tissue (Haemmerich et al 2003). Generally, such differences between carcinoma and normal tissue are attributed to different cellular water content, amount of extracellular fluid, membrane permeability, packing density and orientation of the malignant cells (Zou and Guo 2003). Other than carcinomas, tissues under conditions of ischemia, hemorrhage or edema are expected to exhibit different electrical properties as blood and most body fluid have quite different conductivity and permittivity than most other soft tissues (Fallert et al 1993, Cinca et al 1997). Therefore, noninvasive imaging methods measuring tissue electrical properties with good accuracy and high spatial resolution are of great research and clinical interest.

Over decades, different electromagnetic imaging methods have been developed to measure electrical properties of biological tissue, including electrical impedance tomography (EIT) (Barber and Brown 1984, Cheney et al 1999), magnetic induction tomography (MIT) (Griffiths et al 1999) and magnetic resonance electrical impedance tomography (MREIT) (Khang et al 2002, Woo and Seo 2008). Among these techniques, EIT maps tissue electrical properties using acquired surface voltage measurements in response to different current injections. Though EIT has advantages in its low cost, real-time speed and safety, major limitations including its low spatial resolution and degraded sensitivity in the center of an object still hinder its broader application. In addition, due to its use of current injection through surface electrodes, EIT may be limited by the ‘shielding effect’ (Wen 2000) caused by an insulating or low conductive region in the object, such as bone or adipose tissue. In comparison, MIT uses dynamic magnetic field to induce current in conductive tissue and measures the second magnetic field generated by the induced eddy current using noncontact sensing coils. Yet, because of the ill-posed inverse problem similar to EIT, the spatial resolution of current MIT techniques is still quite limited. In order to achieve high spatial resolution in imaging electrical conductivity, MREIT has been developed by combing EIT and magnetic resonance current density imaging (MRCDI) (Joy et al 1989). With current injection through surface electrodes similar to EIT while measuring the corresponding magnetic field disturbance generated by injected current in tissue through magnetic resonance imaging (MRI), MREIT made it possible to map electrical conductivity in ex vivo and in vivo tissues with high spatial resolution (Woo et al 2008, Seo and Woo 2014). However, a relatively high level of current injection (on the level of mA)
is generally required in MREIT to obtain sufficient signal-to-noise ratio (SNR) level and the use of MRI machines makes the cost of MREIT higher than other methods.

Alternative approaches utilizing the coupling between electromagnetic field and acoustic field have also been developed to image electrical properties of tissue or bioelectrical current (Roth 2011). Such kind of coupling was first demonstrated in magnetoacoustic tomography (MAT) (Towe and Islam 1988, Roth et al 1994) and Hall effect imaging (HEI) (Wen et al 1998). In MAT and HEI, the imaging object is placed in a static magnetic field. Spontaneous or injected current flow, which is associated with ion movement in biological tissue, is then coupled to acoustic vibrations through Lorentz force acting on these moving ions. Such vibrations can be sensitively detected by ultrasound transducers and used for possible mapping of the bioelectric current or tissue electrical properties with spatial resolution close to ultrasound imaging. Using similar coupling mechanism in a reverse mode, one can apply ultrasonic energy to the imaging object and record voltage/current signals to obtain the sample’s conductivity information (Montalibet et al 2001a, 2001b, Roth and Schalte 2009). Such technique was also called magneto-acousto-electrical tomography (MAET) (Haider et al 2008, Renzhiglova et al 2010, Kunyansky 2012) or Lorentz force electrical impedance tomography (LFEIT) (Grasland-Mongrain et al 2013, 2015). Of course, the problem of the ‘shielding effect’ associated with the use of surface electrodes for current injection or voltage measurement, i.e. regions surrounded by low-conductive tissue become invisible, still exists in these methods. Such problems have then led to the development of magnetoacoustic tomography with magnetic induction (MAT-MI) (He 2005, Xu and He 2005). MAT-MI utilizes magnetic induction to induce eddy current in the conductive sample and generates acoustic vibrations through the same Lorentz force coupling mechanism as in MAT or HEI. Ultrasound waves are then sensed to reconstruct the electrical conductivity related image. Ever since the MAT-MI method was proposed, there have been many numerical studies (Li et al 2007, 2009, Ma and He 2007, Zhou et al 2011) and experimental studies using physical phantoms (Li et al 2006, Xia et al 2007, Sun et al 2013) or biological tissues (Hu et al 2011, Hu and He 2011) demonstrating the feasibility and performance of MAT-MI. Advancement on experimental system design (Hu et al 2010, Li and He 2010) and image reconstruction algorithms (Li and He 2010, Mariappan and He 2013) has also been achieved in recent years. In addition, similar to the reverse mode of HEI or MAET, i.e. applying ultrasound transmission and measuring the Lorentz force induced current or voltage for imaging electrical conductivity, the reverse mode of MAT-MI—named magneto-acousto-electrical tomography with magnetic induction (MAET-MI)—has also been developed recently (Guo et al 2015), which uses ultrasound stimulation and coil measurement of the dynamic magnetic field generated by Lorentz force induced current in conductive imaging objects.

In this review, we first go over the basic theories and physical mechanisms underlying the MAT-MI imaging methodology. Different experimental systems and imaging strategies that have been developed during recent years are then discussed and compared. In conclusion we give our opinions on the challenges and possible future directions in MAT-MI research.

2. Theories and physical mechanisms

A schematic diagram of the imaging principle of MAT-MI is shown in figure 1. In MAT-MI, a conductive object with electrical conductivity \( \sigma(\mathbf{r}) \) is placed in a static magnetic field \( \mathbf{B}_0 \) and a time-varying magnetic stimulation with magnetic flux density \( \mathbf{B}_1 \) is applied to induce eddy current \( \mathbf{J} \) inside the object volume \( \Omega \). Note that the induced eddy current is determined by the \( \mathbf{B}_1 \) field and the conductivity distribution \( \sigma(\mathbf{r}) \), where \( \mathbf{r} \) is the position vector. Also note
here that in MAT-MI we are considering around MHz system frequency (central frequency of magnetic stimulation and ultrasound transducer) and conduction current is much larger than displacement current (Xu and He 2005). Therefore, tissue capacitance effect related to permittivity is ignored here and only conductivity of the tissue needs to be considered. Within the static magnetic field, the Lorentz force $\mathbf{F} = \mathbf{J} \times \mathbf{B}_0$ moves those charged ions forming the eddy current and leads to detectable ultrasound pressure signals. These ultrasound signals can be acquired by ultrasound probes and used to estimate the object’s electrical conductivity map.

In the following, theories of the two major physical mechanisms in MAT-MI, i.e. magnetic induction in conductive tissue samples and Lorentz force induced acoustic wave propagation are described.

2.1. Magnetic induction in conductive tissue sample

As in MAT-MI we are considering magnetic stimulation centered around MHz (usually pulsed magnetic stimulation with microsecond pulse duration), the corresponding skin depth of magnetic induction in general biological tissue (assuming conductivity of 0.2 S m$^{-1}$ and relative permeability of 1) is at the level of meters and much larger than most organ size. Therefore, the magnetic induction problem in MAT-MI can be considered quasi-static and magnetic diffusion can be ignored (Li and He 2010). The quasi-static condition allows us to separate the spatial and temporal functions of the time-varying magnetic field, i.e. $\mathbf{B}_i(\mathbf{r}, t) = \mathbf{B}_0(\mathbf{r}) f(t)$. This condition also indicates that the magnetic field in the tissue can be very well approximated by the field produced by the coil in the absence of the tissue and can be estimated with the known coil geometry (Wang and Eisenberg 1994).

Using the magnetic vector potential $\mathbf{A}(\mathbf{r}, t)$ as $\mathbf{B}_i = \nabla \times \mathbf{A}$. According to Faraday’s law, we have

$$\nabla \times \left( \mathbf{E} + \frac{\partial \mathbf{A}}{\partial t} \right) = 0$$

(1)

where $\mathbf{E}(\mathbf{r}, t)$ is the electrical field intensity. Therefore the electrical field intensity $\mathbf{E}(\mathbf{r}, t)$ can be written as:

$$\mathbf{E} = - \nabla \phi - \frac{\partial \mathbf{A}}{\partial t}$$

(2)
where $\phi(r, t)$ is the electrical scalar potential. According to Ampere’s law and because we ignore the displacement current, the current density $J(r, t)$ is solenoidal as:

$$\nabla \cdot J = 0. \quad (3)$$

In addition, according to Ohm’s law, the current density is related to the electrical field through conductivity as:

$$J = \sigma E. \quad (4)$$

Combining equations (2)–(4), we can have:

$$\nabla \cdot (\sigma \nabla \phi) = - \nabla \cdot \left( \frac{\partial \mathbf{A}}{\partial t} \right). \quad (5)$$

According to the quasi-static condition and Faraday’s Law and Ohm’s Law, similar spatial and temporal separation holds for the magnetic vector potential, induced electrical field and eddy current density, i.e. $A(r, t) = A(r) f(t)$, $\phi(r, t) = \phi(r) f'(t)$, $E(r, t) = E(r) f'(t)$ and $J(r, t) = J(r) f'(t)$ where the prime denotes the first order time derivative. Assuming a uniform magnetic stimulation over space, equation (5) has an analytical solution in a two layer concentric spherical model (Li et al 2007), yet for arbitrary geometry, equation (5) must be solved in the whole conductive sample domain $\Omega$ with a Neumann boundary condition on the current density at the outer boundary surface as $J \cdot \mathbf{n} = 0$, where $\mathbf{n}$ is the unit vector norm of the outer boundary $\partial \Omega$. This boundary condition requires the current density component that is normal to the bounding surface to vanish. With such boundary conditions, the final equation describing the magnetic induction problem in MAT-MI (in spatial domain) can be written as

$$\sigma \nabla \cdot \nabla \phi - \nabla \cdot \left( \frac{\partial \mathbf{A}}{\partial t} \right) = 0, \quad (6)$$

If the electrical conductivity is known throughout the whole tissue volume, a unique solution for the electrical potential $\phi$ inside the conductive domain $\Omega$ can be determined up to a reference point. This solution can generally be obtained in arbitrary geometry by using numerical methods such as the finite element method (FEM) (Wang and Eisenberg 1994). With the solution of electrical potential, electrical field and eddy current can be calculated accordingly using equations (2) and (4).

### 2.2. Lorentz force induced acoustic wave propagation

With the magnetically induced eddy current $J$ and the static magnetic field $B_0$, the Lorentz force acting on the eddy current over unit volume can be written as $F = J \times B_0$. Note here that we assumed $B_1 \ll B_0$ as the strength of the dynamic field for magnetic induction is much smaller than that of the static field in most of the MAT-MI experiment systems. According to Newton’s second law of motion and assuming the particle velocity $v$ caused by the Lorentz force is small, we have the following equation (7) (Roth et al 1994, Xu and He 2005):

$$\frac{\partial (\rho_0 v)}{\partial t} = - \nabla p + J \times B_0 \quad (7)$$

where $\rho_0$ is the density of the material at rest and $p$ is acoustic pressure. Taking the divergence of both sides of equation (7), we have equation (8):

$$\frac{\partial (\nabla \cdot (\rho_0 v))}{\partial t} = - \nabla^2 p + \nabla \cdot (J \times B_0). \quad (8)$$
In addition, we have the conservation of mass as in equation (9) and the definition of the adiabatic compressibility of the medium $\beta$ as in equation (10):

$$\nabla \cdot (\rho_0 \mathbf{v}) = -\frac{\partial \rho}{\partial t} \quad (9)$$

$$\beta_s p = \frac{\rho}{\rho_0} \quad (10)$$

where $\rho$ is the density variation. Combining equations (8)–(10) and using the relationship $c_s = \frac{1}{\sqrt{\rho_0 \beta}}$, where $c_s$ is the acoustic speed in the medium, we can derive the wave equation with the Lorentz force induced acoustic source (Roth et al. 1994, Xu and He 2005):

$$\nabla^2 p - \frac{1}{c_s^2} \frac{\partial^2 p}{\partial t^2} = \nabla \cdot (\mathbf{J} \times \mathbf{B}_0). \quad (11)$$

Note here that in MAT-MI the static magnetic field is generally generated from some external sources such as permanent magnets placed outside the imaging object volume, thus $\nabla \times \mathbf{B}_0 = 0$ inside the imaging object volume (Xu and He 2005). The MAT-MI acoustic source, i.e. the right hand side of the wave equation (11), can be written as $AS(r) = \nabla \cdot (\mathbf{J} \times \mathbf{B}_0) = (\nabla \times \mathbf{J}) \cdot \mathbf{B}_0$. Note here that since the MAT-MI acoustic source is related directly to the curl of the eddy current density, the irrotational part of the current density does not contribute to MAT-MI acoustic sources (Mariappan and He 2013). According to both Ohm’s law and Faraday’s law and by using the quasi-static condition to separate the spatial and temporal functions, the MAT-MI acoustic source $AS(r, t)$ can be further expanded as

$$AS(r, t) = AS(r)^f(t) = (\sigma(\nabla \times \mathbf{E}) + \nabla \sigma \times \mathbf{E}) \cdot \mathbf{B}_0$$

$$= \left(-\sigma \frac{\partial \mathbf{B}_0}{\partial t} + \nabla \sigma \times \mathbf{E}\right) \cdot \mathbf{B}_0 = (-\sigma \mathbf{B}_0(r) + \nabla \sigma \times \mathbf{E}(r)) \cdot \mathbf{B}_0 \cdot f'(t). \quad (12)$$

As shown in equation (12), besides the static and dynamic magnetic fields, the MAT-MI acoustic source is related to both the conductivity and its spatial gradient. Assuming the medium is acoustically homogeneous and the acoustic speed $c_s$ is a constant over space, using the 3D Green’s function, the solution to equation (11) can be written as (Xu and He 2005):

$$p(r_0, t) = -\frac{1}{4\pi} \int \int_V \mathbf{dr} \cdot AS(r) \cdot \frac{\delta(t - |r_0 - r|/c_s)}{|r_0 - r|} \quad (13)$$

where $r_0$ is a position located on a certain ultrasound detection aperture.

In equation (13) we assumed the Lorentz force induced acoustic source does not have specific propagation direction. In comparison, some previous studies modeled such Lorentz force induced acoustic source as acoustic dipole radiations (Ma and He 2008, Sun et al. 2012, 2013, Wang et al. 2014), i.e. acoustic sources with specific propagation direction that is aligned with the Lorentz force. Such dipole acoustic radiation has been modeled using equation (13) with an extra spatial derivative on the acoustic source term in the direction of the Lorentz force (Ma and He 2008, Sun et al. 2012) or with an extra projection term—i.e. $\cos \theta$, where $\theta$ represents the angle between the direction of the Lorentz force $\mathbf{F}$ and the transmission line between acoustic source position and the plane transducer $r_0 - r$ (Sun et al. 2013, Wang et al. 2014). The directional acoustic source in MAT-MI has been previously demonstrated in experiment with line-shape objects (Wang et al. 2014), yet due to the directional nature of most piston
transducers, similar directional behavior may also have been observed with non-dipole acoustic radiations (Mariappan et al 2011), i.e. acoustic sources residing on tissue boundaries that are perpendicular to the transducer’s surface cannot be effectively detected. Therefore, further experimental validation may be needed to prove that in MAT-MI the Lorentz force does induce acoustic dipole radiations.

2.3. Reverse mode—MAET-MI

As a reverse model of MAT-MI, MAET-MI (Guo et al 2015) was recently developed to map electrical conductivity using ultrasound stimulation and coil detection of the magnetic field generated by the Lorentz force induced current.

In the forward problem of MAET-MI, two sources that contribute to the coil-detectable magnetic field $H_1$ are the source current density $J_1$ and the corresponding electric field $E_1$. If the object $\Omega$ has a conductivity distribution $\sigma$, the relations among these fields can be described as a first governing equation set:

$$\begin{align*}
\nabla \times H_1 &= J_1 + \sigma E_1 \\
\nabla \times E_1 &= -\mu \frac{\partial H_1}{\partial t} \\
J_1 &= \sigma (v \times B_0)
\end{align*}$$

in which $\mu$ is the permeability of the medium and $v$ is the velocity of local conductive particles driven by ultrasound field. In order to determine the induced voltage $u$ across the detection coil, reciprocal theorem is applied to obtain a second equation set:

$$\begin{align*}
\nabla \times H_2 &= J_2 + \sigma E_2 \\
\nabla \times E_2 &= -\mu \frac{\partial H_2}{\partial t} \\
J_2 &= \sigma E_2
\end{align*}$$

when an external current $J_2$ is injected into the coil, inducing eddy current $J_2$ within $\Omega$. In fact, such reciprocal process of MAET-MI is similar to the actual working process of MAT-MI discussed above, in which magnetic induction is applied to produce an electric field $E_2$ over the object. Then, both $J_2$ and $E_2$ contribute together and relate to the magnetic field $H_2$.

Based on equations (14) and (15), one can solve for the voltage $u$ by taking the assumptions of infinite boundary condition and non-conductive property at the boundaries of $\Omega$. This mathematical procedure can be simplified to equation (16), as long as the spectral content of $J_2$ is non-zero within the spectrum range of the measured voltage $u$:

$$u(t) = \int \int_{r \in \Omega} \frac{B_0}{\rho} \cdot \nabla \times J_2(r) \cdot \varphi(r, t) \, dr$$

in which $\rho$ is the mass density, and $\varphi(r, t)$ is the velocity potential due to the ultrasound vibration. This velocity potential satisfies $\nabla \varphi = \rho v$, and can be further determined from the wave equation and the corresponding Green’s function by taking acoustic far-field assumption. Another unknown variable needed to solve equation (16) is $J_2(r)$, the spatial component of the induced eddy current $J_2$. Approaches to calculating such a vector field have been proposed in MAT-MI (Li et al 2007, Li and He 2010, Mariappan and He 2013), and an appropriate method can be chosen depending on the specific geometric feature and conductivity distribution of the object $\Omega$. 

3. Experimental setups

According to the physical processes involved, a MAT-MI experimental system has three major components: a static magnetic field, a dynamic magnetic field for current induction, and a scanning system with ultrasound transducer for acquiring ultrasound signals. A typical experimental setup for MAT-MI imaging systems is illustrated in figure 2. Each component of the MAT-MI experimental system is discussed below.

The static magnetic field in MAT-MI is usually generated by permanent magnets, which can give field strength of 0.1–0.3 T around the sample in z direction (Xia et al 2007, Hu et al 2011). In a recent study, an MAT-MI system using a superconducting magnet in a 9.4T MRI scanner have also been explored (Mariappan et al 2014) with a slightly different scanning framework, i.e. due to the limited space in the MRI bore, the transducer and coil were fixed while rotation of the imaging sample was performed.

To generate dynamic magnetic stimulation, computer controlled magnetic stimulators are used to drive coil load arranged around the imaging sample. For MAT-MI with pulsed magnetic stimulations, the magnetic stimulators usually use high voltage and high current switches to control capacitor discharge through connected coil load. Depending on the hardware design, high voltage capacitors that charge to 600V (Xia et al 2007) and to as high as 24 kV have been used in previous MAT-MI systems (Hu et al 2010). The corresponding maximum dynamic magnetic field, which can be measured by small sensing coils around the imaging object, ranged from 0.001 25 T to 0.07 T (Xu and He 2005, Hu et al 2010, Li and He 2010). Magnetic stimulations sent though different coils in MAT-MI have also been explored in the past. Most MAT-MI systems use a single magnetic stimulation generated through one coil set, such as a Helmholtz coil pair, which produces a region with a nearly uniform magnetic field (Xia et al 2007, Mariappan and He 2013). In addition, MAT-MI systems using different coil sets to generate different magnetic stimulations and consequently different eddy current patterns in the conductive tissue sample have also been developed in the so-called multi-excitation MAT-MI system (Li and He 2010).

Ultrasound transducers mounted to certain scanning systems that can either rotate the transducer or the imaging sample can be used to acquire MAT-MI signals around the imaging sample. Usually both the sample and transducer are submerged in acoustic coupling media with very low electrical conductivity, e.g. distilled water. Data acquisition is synchronized with the magnetic stimulation. Signals detected by the transducer are then filtered, amplified and fed to the data acquisition system. In MAT-MI, the magnetic induced eddy current in conductive tissues and the Lorentz force induced acoustic source are intrinsically distributed in the three-dimensional (3D) tissue volume. Thus 3D scanning and mapping of the Lorentz force induced acoustic source is needed to map the 3D electrical conductivity distribution. 3D MAT-MI systems have been previously developed based on ultrasound focusing and cylindrical scanning (Xia et al 2007, Li et al 2010). As shown in figure 2, through ultrasound focusing in the Z direction, we can localize the MAT-MI acoustic sources in a specific XY plane. A 2-dimensional (2D) MAT-MI image can be obtained at each cross section of the 3D object and vertical scans in the Z direction can provide a stack of 2D images, thus forming a 3D volume image of the object. With such a circular scan, the spatial resolution of the MAT-MI image in each slice is mainly determined by the central frequency and bandwidth of the transducer, while the resolution in the Z direction is determined by the focusing beam width. For imaging objects that are uniform in the Z direction and with homogeneous static and dynamic magnetic field, the whole MAT-MI system can be simplified as a 2D system, in which only 2D ultrasound scan is needed (Li et al 2006, Hu et al 2010). For 2D ultrasound scans, besides the circular scan mode with large scanning radius around the objects using
unfocused transducers, scanning in B-mode scan using focused transducers may also be used (Mariappan et al 2011).

As the reverse mode of MAT-MI, MAET-MI also includes three major components, a static magnetic field, an ultrasound stimulation system with waveform control, and a detecting system using detection coils close to the object. The static magnetic fields in MAET-MI systems are also created by permanent magnets, and the applied field strength ranges from 0.2–0.3 T around the imaged object along the Z direction (Guo et al 2015). Depending on system design, the ultrasound stimulation may employ transducers working at different center frequencies, e.g. 500 kHz with a pulse repetition frequency (PRF) of 2 kHz (Guo et al 2015). Similar to the high voltage used to instantaneously excite the radio-frequency coil in MAT-MI, an 80 ns pulsed, 1200 V amplitude voltage was applied in a recent MAET-MI study (Guo et al 2015) to drive the ultrasound transducer. It is believed that optimized ultrasonic parameters could lead to a sufficient acoustic pressure onto the region of interest (ROI), thus producing significant mechanical vibrations (the velocity) to potentially achieve a good SNR for the acquired MAET-MI data. In addition, higher center frequencies and smaller oscillation numbers of the transmitted ultrasound can result in better spatial resolutions. In that study, two 150-turn coils connected in series were arranged and fixed at the top and bottom sides of the object. The acquired signals were then fed to a low noise amplifier, and were further averaged 1024 times to improve SNR (Guo et al 2015). MAET-MI faces a practical challenge in that the ultrasonic wave may also vibrate the sensing coils in detection, thus raising significant artifacts in the reconstructed image. Therefore, the distance between the sensing elements and the ultrasound targeted ROI needs to be optimized, such that the artifacts in the collected signal can be suppressed, while the detecting sensitivity will not be sacrificed too much (Guo et al 2015). Such problem is of course mainly due to the whole experimental setup being immersed in nonconductive oil in that study, which potentially provides acoustic pathways between the coil and ultrasound transducer. One good practice arising from MAET-MI experiment is its shielding design using an aluminum tank, which is believed to improve the signal quality. To our best knowledge, current MAET-MI images are 2D based, and 3D implementation of MAET-MI with focused ultrasound technique may be developed in the future.

Figure 2. Diagram of an example setup of a MAT-MI experiment system. This figure was reprinted with permission from Xia et al (2007).
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4. Imaging strategies and experimental results

The goal of MAT-MI is to reconstruct the electrical conductivity distribution $\sigma(r)$ inside the tissue volume, with the knowledge of the static magnetic field $B_0(r)$, the dynamic magnetic stimulation $B_1(r, t)$ or its vector potential $A(r, t)$ and the measured MAT-MI acoustic signals $p(r_0, t)$. As discussed above, the signal generation mechanism of MAT-MI includes both the processes of magnetic induction and acoustic wave propagation with the Lorenz force induced acoustic vibrations. Accordingly solving the inverse problem of MAT-MI often involves two steps. In the first step, we reconstruct certain intermediate variables related to the volume conductivity from the measured acoustic signals. This may mean the Lorentz force induced acoustic sources (Xu and He 2005, Li et al 2007, Li and He 2010), the Lorentz force (Xia et al 2009, 2010) or the eddy current (Mariappan and He 2013, Mariappan et al 2014). In the second step, we then reconstruct the conductivity distribution from the intermediate variable we get from the first step. It should be noted that the inverse problem of MAT-MI is not as ill-posed as in EIT or MIT techniques because the ultrasound measurements collected around the sample can be used to estimate the selected intermediate variable in each pixel/voxel in the whole imaging volume. This is because the acoustic sources over space are time resolved in the measured acoustic signal due to their acoustic time of flight difference. In comparison, the electromagnetic measurements used in EIT or MIT at each location are always a volume integration of the product between the current density and the lead field of the probes, e.g. surface electrode or coil.

4.1. Mapping conductivity through mapping MAT-MI acoustic source

For simplicity, we often assume a pulsed magnetic stimulation and letting $f'(t) = \delta(t)$ and assume the medium is acoustically homogeneous, i.e. the acoustic speed $c_s$ is a constant in the tissue volume. Applying the time reversal method in MAT-MI (Xu and Wang 2004, Xu and He 2005), the MAT-MI acoustic source can be estimated as

$$\text{AS}(r) \approx \frac{-1}{2\pi c^3_s} \int_{S_0} dS_0 \frac{n_0 \cdot (r-r_0)}{|r-r_0|^2} \left. \frac{\partial^2 p(r_0, t)}{\partial r^2} \right|_{r = |r-r_0|}$$.  \hspace{1cm} (17)

where $r_0$ is a point on the detection surface $S_0$ and $n_0$ is a unit vector normal to the surface $S_0$ at $r_0$.

This is the most commonly used method for reconstructing MAT-MI acoustic sources. Experimentally, the MAT-MI acoustics sources have been shown to be strongest at tissue boundaries with large electrical conductivity change (Li et al 2006) and proportional to the magnitude of conductivity gradient (Li et al 2006, Hu et al 2010), i.e. the higher the gradient magnitude the stronger the detected ultrasound signal and reconstructed MAT-MI acoustic source. This is not totally unexpected because theoretically the MAT-MI acoustic source is related to both the conductivity and its spatial gradient as shown in equation (12). Recent simulations have shown that the conductivity gradient related acoustic source $(\nabla \sigma \times \mathbf{E}(r)) \cdot \mathbf{B}_0$ will be at least five times larger than its counterpart related to conductivity itself, i.e. $\sigma \mathbf{B}_1(r) \cdot \mathbf{B}_0$, unless the transition zone between pieces with different conductivity values is wider than 10% of the piece size (Wang et al 2016). In addition, the bandwidth limitation of most transducers used in MAT-MI experiments will further filter the acoustic signals generated by these two sources, most possibly favoring the gradient source which is a wide-band source in piecewise homogeneous samples (Li and He 2010).

Though it is not a direct measure of tissue electrical conductivity, the MAT-MI acoustic source itself—reconstructed using equation (17)—can give some valuable information in
regards to the boundaries between different tissue types with different electrical conductivities, e.g. muscle versus fat tissue. In recent MAT-MI studies using a static magnetic field of 0.2–0.3 T, magnetic stimulator equipped with high voltage discharging system (with capacitors charged up to 24 kV that can give maximum magnetic field strength $B_0$ of 0.07T around the imaging object), it has been demonstrated that MAT-MI acoustic sources can be detected between ex vivo fat tissue (with conductivity of 0.02–0.03 S m$^{-1}$) and muscle tissue (with conductivity of 0.55–0.62 S m$^{-1}$) (Hu et al 2010). As shown in figures 3(c) and (d), such a system has also been shown able to detect MAT-MI acoustic sources generated at the tissue boundary between liver tumor (with conductivity of 0.65–0.7 S m$^{-1}$) and normal liver tissue (with conductivity of 0.25–0.28 S m$^{-1}$) (Hu et al 2011).

With the reconstructed MAT-MI acoustic sources, there are several ways to further calculate the electrical conductivity. Early approaches assumed piecewise homogeneous conductivity distribution and ignored the conductivity gradient sources (Xu and He 2005, Li et al 2007). Though demonstrated in computer simulation, such simplified approaches can hardly be applied on experimentally collected MAT-MI signal, which is dominated by conductivity gradient sources. As mentioned in some later theoretical studies (Kunyansky 2012, Zhou et al 2014), if it is possible to rotate the static magnetic field $B_0$ e.g. setting it to three orthogonal directions, while keeping the magnetically induced eddy current in the conductive tissue sample the same, one may be able to get a good estimation of the curl of the eddy current $\nabla \times J$ and further estimate the electrical conductivity. However, rotation of the static magnetic field is usually hard to manage in practice, especially when acoustic pathway and ultrasound transducers need to be arranged around the imaging sample. Reconstructing the electrical conductivity from the estimated MAT-MI acoustic sources can be done without rotating the static magnetic field component in $U$ depends on the unknown conductivity distribution $\sigma(r)$ and the

\[ U = \begin{bmatrix} E_y & -E_x \\ \vdots & \vdots \\ E_y & -E_x \\ \end{bmatrix}, \quad x = \begin{bmatrix} \frac{\partial \sigma}{\partial x} \\ \frac{\partial \sigma}{\partial y} \end{bmatrix} \quad \text{and} \quad b = \begin{bmatrix} \frac{\partial^2 \sigma}{\partial x^2} + \sigma B_{1z} \\ \vdots \\ \frac{\partial^2 \sigma}{\partial y^2} + \sigma B_{1z} \end{bmatrix} \]

Such a matrix equation however needs to be solved in an iterative way, as the electrical field component in $U$ depends on the unknown conductivity distribution $\sigma(r)$ and the
vector \( \mathbf{b} \) contains a term related to the conductivity distribution. In order to compute \( \sigma \) from
\[
\nabla \sigma = \left( \frac{\partial \sigma}{\partial x}, \frac{\partial \sigma}{\partial y} \right)
\]
in all the imaging slices, a 2D layer potential integration technique can be used as in (Oh et al 2003)

\[
\sigma(\mathbf{r}) = -\int \nabla_{\mathbf{r}'} \Phi(\mathbf{r} - \mathbf{r}') \cdot \nabla \sigma(\mathbf{r}') d\mathbf{r}' + \int_{\partial S} \mathbf{n}_{\mathbf{r}'} \cdot \nabla_{\mathbf{r}'} \Phi(\mathbf{r} - \mathbf{r}') \sigma_{\partial S}(\mathbf{r}') d\mathbf{r}'
\]

(20)

where \( \Phi(\mathbf{r} - \mathbf{r}') = \frac{1}{2 \pi} \log|\mathbf{r} - \mathbf{r}'| \) is the 2D Green’s function of the Laplacian operator and
\( \nabla_{\mathbf{r}'} \Phi(\mathbf{r} - \mathbf{r}') = -\frac{1}{2 \pi} \frac{\mathbf{r} - \mathbf{r}'}{|\mathbf{r} - \mathbf{r}'|^3} \). \( S \) denotes the 2D imaging ROI in the imaging slice where \( \nabla \sigma \) is obtained and \( \partial S \) denotes its boundary. \( \sigma_{\partial S} \) is the conductivity value restricted at the boundary \( \partial S \). The 2D integration as in equation (20) can be applied in a whole 3D volume slice by slice. In addition, modifications can be made on equations (18) and (19), i.e. removing the terms
related to conductivity $\sigma$ while keeping those terms related to conductivity gradient $\nabla \sigma$, to account for the bandwidth limitation in the ultrasound measurement (Li and He 2010). Such modification was based on the fact that with limited bandwidth acoustic measurements, the reconstructed acoustic source is mainly determined by the conductivity gradient term. Similar iterative procedure can be used to solve the inverse problem. However, as shown in both computer simulation and experiment studies, using the multi-excitation gradient approach with limited bandwidth acoustic measurements, we are only able to reconstruct the relative conductivity contrast (Li and He 2010, Li et al 2010).

Some example results acquired in a phantom using a 2D multi-excitation MAT-MI system are shown in figure 4. In that experiment, the static magnetic field was measured to be 0.26 T (Gaussmeter, Alpha Lab) at the coordinate center where the object was located. Three coil sets were used to send three different magnetic excitation patterns, including a Helmholtz coil pair (group C in figure 4(a)) and two figure-eight coil pairs (group A and B in figure 4(a)). The distance between the upper coils and lower coils in each group was around 5 cm. The coils were driven by 1 $\mu$s current pulse. The estimated maximum dynamic magnetic field strength $B_{1z}$ was around 0.007 T at the coordinate center. A 500 kHz flat ultrasound transducer (Panametrics V301) with around 60% bandwidth was used to scan around the sample with 330 degrees view angle and 2.5 degrees step size. The scanning radius, i.e. the distance between the transducer and the scanning center was 22.8 cm. A 3 cm thick gel phantom (figure 4(b)), which was uniform in the Z direction was submerged in 3 cm thick deionized water medium for acoustic coupling. The phantom contained a background region made from 5% salinity gel ($\sim$8 S m$^{-1}$). Two cylindrical columns with diameter of 12 mm were embedded in the gel. Marked by the red and blue circles in the photo are two high conductive regions filled with 20% ($\sim$22 S m$^{-1}$) and 10% ($\sim$13 S m$^{-1}$) salinity gels respectively. These two regions had diameter of 8 mm. The two annular areas sitting between the two high conductive regions and the background were made from beef suet ($\sim$0.03 S m$^{-1}$). The acoustic signal collected using the transducer was fed into preamplifiers with 90 dB gain and digitized by a 5 MHz data acquisition card.

The acoustic source images reconstructed using equation (17) corresponding to the three different magnetic excitations (group A, B and C in figure 4(a)) were shown in figures 4(c)–(e) respectively. Different acoustic source distributions under different magnetic stimulations were observed, which agrees with the predicted pattern using computer simulation (Li and He 2010). Using the modified multi-excitation algorithm, the conductivity image of the gel phantom was reconstructed as shown in figure 4(f). From this image, we can clearly see the relative conductivity contrast, while the fat layer shows lower conductivity than the surrounding background, the 10% salinity gel shows higher conductivity value and the 20% salinity gel shows the highest conductivity. A conductivity profile at $y = 0.01$ m is given in figure 4(g) showing the comparison between the target and reconstructed conductivity values.

4.2. Mapping conductivity through reconstructing vector sources

Previous theoretical studies have shown that if MAT-MI ultrasound signal can be collected at specific acoustic apertures, e.g. spherical aperture or cylindrical aperture, at large distance from the source field (far field assumption), the Lorentz force vector $\mathbf{F} = \mathbf{J} \times \mathbf{B}_0$ may be reconstructed by time-reversing vectorized acoustic pressure measurements (Xia et al 2009, 2010). Electrical conductivity may be further estimated given the knowledge of both the static and the dynamic magnetic fields, i.e. $\mathbf{B}_0$ and $\mathbf{B}_1$. Though such methods have been demonstrated theoretically using computer simulations, it is hard to validate them in experiment due to the
Figure 4. (a) Diagram of three different coil setups, i.e. group A, B and C, used in the multi-excitation MAT-MI system. Different coil setups were designed to induce different eddy-current patterns in the imaging object. (b) A gel phantom used to test the multi-excitation MAT-MI method. The red circle marks a region containing 20% salinity gel and the blue circle marks a region containing 10% salinity gel. (c)–(e) Reconstructed MAT-MI acoustic source images under magnetic excitations delivered through the coil groups of A, B and C, respectively. (f) Reconstructed conductivity image showing the relative conductivity contrast. (g) Conductivity profile along $y = 0.01$ m showing the comparison between the target and reconstructed conductivity values. This figure was reproduced with permission using sources from Li and He (2010).
requirement of acoustic measurements on special 3D acoustic apertures. Nevertheless, they gave some new thoughts about how to estimate the electrical conductivity in MAT-MI.

Inspired by the Lorentz force mapping method, recently an ultrasound beam forming method has been developed for MAT-MI to map the electrical conductivity through the estimation of magnetic induced eddy current vector (Mariappan and He 2013, Mariappan et al 2014). We call it current density vector source method here. The basic idea is to design a certain point spread function (psf) of the ultrasound imaging system to extract the orthogonal components of the current density. The beam-forming algorithm comprises summing up the weighted and time delayed pressure signal at all the receiver locations to synthesize a certain signal source at the source location \( r_s \) as

\[
V(r_s) = \sum_n W_n p(r_n, t - t_a)
\]

(21)

where \( W_n \) is the weight and \( t_a \) is the time delay applied to the signal; \( n \) is the number of ultrasound receiver used in the beam formation process. Substituting equations (12) and (13) in equation (21) and applying the fact that the MAT-MI acoustic source vanishes outside the conductive region (Xia et al 2009, Mariappan and He 2013), we get

\[
V(r_s) = \sum_n W_n \int \int \int_V \nabla \cdot (J \times B_0) \cdot \nabla G(r_n, r, t - t_a)
\]

(22)

where \( G \) is the Green’s function for the wave equation, i.e. \( G(r_n, r, t) = \frac{\delta(t - |r_n - r|)}{|r_n - r|} \).

Switching the integration over space and the summation in beam forming process and using the vector identity for the triple vector product, equation (22) can be rewritten as

\[
V(r_s) = \int \int \int_V \nabla \cdot J \left( B_0 \times \sum_n W_n \nabla G(r_n, r, t - t_a) \right)
\]

(23)

With appropriately designed weights and time delays, the vector psf in equation (23), i.e. \( S(r) = B_0 \times \sum_n W_n \nabla G(r_n, r, t - t_a) \) can be constructed to be a spatial impulse vector pointing to one of the orthogonal directions, e.g. making \( S(r) = \delta(r) \hat{x} \) or \( S(r) = \delta(r) \hat{y} \); assuming \( B_0 = B_0 \hat{z} \), corresponding components of the current density \( J_x \) and \( J_y \) can then be estimated. Since the irrotational part of eddy current does not contribute to the MAT-MI acoustic source, in a piecewise homogenous sample, the conductivity in any homogeneous piece can be estimated by (Mariappan and He 2013)

\[
\sigma \approx -J_z \left( \frac{\partial A}{\partial t} \right)
\]

(24)

where \( J_z \) is the rotational part of the induced eddy current in homogeneous regions. Note here that equation (24) does not apply at conductivity boundaries. Inside each homogeneous region, the conductivity value can be robustly estimated using a least square fit (Mariappan and He 2013). In recent studies on this vector source reconstruction method (Mariappan and He 2013, Mariappan et al 2014), the bandwidth limitation of ultrasound measurement was handled with inverse filtering and least square fitting over space in homogeneous regions, while the conductivity values at or near conductivity boundaries were calculated by interpolation. Computer simulations and experimental studies have been conducted to demonstrate the validity of this method.

Some experimental results acquired in a gel phantom using this method are demonstrated in figure 5. In such an experiment, static magnetic field was measured to be 0.2 T and a Helmholtz coil pair was used to generate pulsed magnetic field (a bipolar single cycle sinusoid
Figure 5. (a) Gel phantom used to test MAT-MI with beam forming vector source reconstruction. (b) MAT-MI acoustic source image of the phantom. (c)–(d) The reconstructed x and y components of the induced eddy current in the conductive sample, respectively. (e) The reconstructed conductivity image of the phantom. (f) Line profile at x = −1 mm showing the comparison between the target and reconstructed conductivity values. This figure was reprinted with permission from Mariappan and He (2013).
with 2 µs pulse width) around 0.006 T at the coordinate center. The MAT-MI acoustic signal measurement was conducted using a circular scanning scheme with a 500kHz flat transducer (Panametrics V301) and around 60% bandwidth. Again because of the uniform conductivity distribution of the imaging sample and the relatively uniform magnetic field in the Z direction, the corresponding MAT-MI problem can be simplified as a 2D problem. The acoustic signals collected were pre-amplified with 90 dB gain and digitized at 5 MHz sampling. Band-pass filtering with passband of 100 kHz to 900kHz was applied to remove electromagnetic interference (EMI) noises. After each experiment, the conductivity of the sample was measured using a four-electrode probe (Hu et al 2010). The reconstructed conductivity distribution using the current density vector source method was then normalized to the expected range of values using a calibration factor to account for various gains in the system. The phantom was made with a 0.4% salinity gel (0.67 S m\(^{-1}\)) in the background and a 2.5 cm diameter cylindrical column with 1.2% salinity gel (2.01 S m\(^{-1}\)) in the center (the black part in figure 5(a)). The reconstructed acoustic source distribution is shown in figure 5(b). As expected, with limited bandwidth measurement the MAT-MI acoustic source was mainly distributed around the conductivity boundaries in the sample. The rotational nature of the induced current was observed in the two components of the reconstructed current density, i.e. \(J_x\) and \(J_y\) (figures 5(c) and (d)). The estimated conductivity distribution (figure 5(e)) and quantitative values (as shown in the profile in figure 5(f)) agreed well with the expected values in the phantom.

4.3. Inverse problem of MAET-MI

To solve the inverse problem in MAET-MI (Guo et al 2015), a compressed sensing method was introduced to solve an intermediate variable \(D(r)\), which is a distributed source function for a constructed wave. Further, a matrix form of equation (16), i.e. \(U = \Phi D\) (a constructed vector of \(D(r)\)), and empirical observations of the sparsity of \(D\) in a certain functional basis \(\Psi\), i.e. \(D = \Psi \theta\), enables one to transform the problem of reconstructing \(D(r)\) to a \(l_1\) norm minimization problem:

\[
\begin{aligned}
\min & \|\theta\|_1 \\
\text{subject to} & \|U - \Phi \Psi \theta\|_2 < \varepsilon
\end{aligned}
\]

in which \(U\) is a vector of the voltage signal detected using the coil in the forward problem, and \(\Phi\) is a known matrix composed of time derivatives of \(\varphi(r,t)\) at discretized time points and spatial locations. However, to use equation (25), MAET-MI requires \(\Phi \Psi\) to be a compressed-sensing matrix, which needs \(\varphi(r,t)\) to be a linear independent function. This condition implies that ultrasound transmissions, i.e. the distance between a transducer and the imaged object and the angular step of the transducer in scan, need to be carefully designed.

The next step is reconstructing the conductivity distribution \(\sigma\) from the calculated \(D(r)\). Based on the assumption of uniform \(B_0\) and \(\rho\), the spatial component of \(\nabla \times J_3(r)\) along the same direction of \(B_0\) can thus be determined from the distributed source function \(D(r)\), which is denoted as \(P\). By further using the hypothesis that the conductivity at the boundaries of \(\Omega\) vanishes, i.e. \(\sigma_{\Gamma \in \partial \Omega} = 0\), a discretized matrix form can be constructed as:

\[
C \sigma = -P
\]

in which \(C\) is a coefficients matrix varying with \(\sigma\), and thus the reconstruction of \(\sigma\) is an iterative process. The Levenberg–Marquardt algorithm can be used to solve this non-linear least squares problem, thus obtaining the correction value for \(\sigma\) in each iterative step while keeping both matrices \(C\) and \(\sigma\) updated.
5. Challenges and future directions

In spite of many technical advances achieved in recent years on MAT-MI, limitations still exist in currently available MAT-MI methods in the aspects of sensitivity, instrumentation and reconstruction algorithms and further improvement is necessary to make it useful and applicable in clinical settings.

As discussed before, the strength of MAT-MI acoustic source is related to the static magnetic field \( B_0 \), the dynamic magnetic stimulation \( B_1 \) and electrical conductivity of the imaging sample. For the ultimate application goal of MAT-MI, e.g. for tumor detection, a sensitivity of detecting electrical conductivity contrast on the level of 0.01 S m\(^{-1}\) is necessary. Current MAT-MI imaging systems are getting close to such sensitivity, but further improvement may be achieved by pushing the static magnetic field strength and dynamic stimulation. First, in most MAT-MI experimental systems developed in previous studies, due to cost and simplicity, permanent magnets were used to give a static magnetic field up to 0.2 to 0.3 T around the imaging sample. Obviously the strength of the static magnetic field in MAT-MI has much room for improvement, e.g. the signal strength may increase more than 10 times in clinically available 3T or even ultrahigh field (7 T and above) MRI scanners. The increase of MAT-MI signal at higher static magnetic field has been recently observed at 9.4 T (about 14 times increase in signal strength and SNR) (Mariappan et al 2014), yet technical challenges in terms of developing a MAT-MI system that works in an MRI machine have also been observed. Such challenges partly come from the need to use magnetic stimulations with short pulses and relatively low central frequency (around MHz) in MAT-MI as compared to the Larmor frequency in MRI (400 MHz at 9.4 T), thus MRI coils cannot be used directly for MAT-MI. In addition, to comply with MRI safety regulations, long cables (~7 m) had to be used to connect the coils and transducers with the magnetic stimulators and amplifiers outside the MRI shielding room, thus energy loss and noise pickup in the long cable were not trivial and partly cancelled out the SNR gains obtained by using the high field (Mariappan et al 2014). Better-designed instrumentation will have to be developed in the future to fully appreciate the SNR gains obtained by doing MAT-MI in high field MRI. Second, with the use of solid-state high voltage and high current switching, the dynamic magnetic stimulation used in MAT-MI system has been substantially improved to about 0.07 T in its maximum dynamic range, which corresponds to a stimulating electric field of about 550 V m\(^{-1}\) (Hu et al 2010, 2011). This is comparable to the magnetic stimulation strength used in transcranial magnetic stimulation (TMS) but still lower than the nerve stimulation threshold of \( 2 \times 10^{-3} \) V s m\(^{-1}\) given the \( \mu \)s pulse duration (Wen et al 1998). So there may be some room for improvement in terms of the magnetic stimulation. Using the magnetic stimulation strength of the latest MAT-MI system (0.07 T) and about 0.2–0.3 T static magnetic field, previous studies have shown that MAT-MI signals can be detected at tissue boundaries with conductivity contrast at about 0.03 S m\(^{-1}\) (Hu et al 2010, 2011) between tumorous tissue and normal tissue. Note here that with the increased magnetic stimulation strength, the assumption \( B_1 \ll B_0 \) used for equation (7) may need to be modified to include the contribution of \( J \times B_1 \) in the induced Lorentz Force. With improved field strength and stimulation, sensitivity to detect 0.01 S m\(^{-1}\) conductivity contrast using MAT-MI is believed to be feasible. In addition, the improved sensitivity and SNR gain would also allow much less signal averaging and speed up the MAT-MI scan. MAT-MI imaging speed may be further improved by using electrically controlled ultrasound phased arrays instead of single element transducers and mechanical scanning.

Another technical challenge in MAT-MI instrumentation is the EMI caused by the time varying magnetic field on the ultrasound transducer. Related to the turn off transients in the stimulator and the impulse response of the transducer, such EMI usually contaminates the
MAT-MI signal for substantial long period (tens of $\mu$s) and have observed some low frequency variations (Mariappan et al 2014). Therefore, further band-pass filtering is usually needed before MAT-MI image reconstruction. Better ways to avoid the EM contamination to ultrasound transducer, such as better EM shielding or the use of optical fiber based transducers, may be further explored to allow acquiring ultrasound signals in a closer distance (e.g. when using transducer with shorter focus) and improve the MAT-MI signal quality.

In terms of reconstruction algorithms, for electrical conductivity reconstruction through mapping MAT-MI acoustic sources such as the multi-excitation MAT-MI method, more sophisticated coil pattern designs may help improve the inverse conditions and the final image quality. On the other hand, the recently developed beam-forming current density vector source method has shown great potential for electrical conductivity reconstruction with the use of only single magnetic excitation. Ultrasound bandwidth limitations have been successfully tackled using inverse filtering in that method. Further improvement may be achieved by considering the conductivity gradient term using iterative processes when calculating the electrical conductivity from the estimated current density vector.

In most previous MAT-MI studies, homogeneous acoustic properties and isotropic electrical conductivity in the imaging sample have been assumed. This is however not always true in biological tissues (Duck 1990, Gabriel et al 1996). Small variations in acoustic properties including acoustic speed and tissue density or acoustic impedance (though less than 10%) in soft tissue is an important source for ultrasound speckle patterns and have been observed in Lorentz force electrical impedance imaging (Grasland-Mongrain et al 2015). Such inhomogeneity may also partly explain some textures observed in previous MAT-MI images acquired in tissue samples, e.g. in the tumor tissue in figure 3(d). In addition, such heterogeneity may cause imaging artifacts, as it was not accurately modeled in most current MAT-MI theory. Acoustic reflections due to the acoustic impedance heterogeneity will also create extra ‘noisy’ acoustic sources that are not related to the electrical properties of the imaging objects and may provide extra information of the mechanical properties if modeled correctly in the future. Some recent studies have started to tackle this problem, and initial theoretical and computer simulation studies have demonstrated the feasibility to better quantifying the electrical conductivity in MAT-MI by mapping the inhomogeneous acoustic speed from ultrasound transmission tomography (Zhou et al 2014). In addition, MAT-MI image reconstruction algorithms that can take into account conductivity anisotropy need to be developed in the future, as electrical conductivity anisotropy in certain tissue types such as muscles and neural fibers has been suggested to change the MAT-MI signal significantly (Brinker and Roth 2008, Li et al 2013).

Theoretically it is also possible to image the Lorentz force induced MAT-MI acoustic sources using the corresponding signal due to shear wave propagation, which would include extra useful information related to tissue shear modulus and stiffness etc. However, due to the much larger attenuation of shear wave than acoustic longitudinal wave in normal soft tissue (several orders of magnitude difference), sensitivity is an issue when measuring the shear wave directly. In a recent study (Grasland-Mongrain et al 2014), Lorentz force induced tissue displacement on the order of 1 $\mu$m was successfully detected by ultrasound speckle tracking. This provides another possible path for imaging both the electrical and elastic properties of tissue using shear wave, which may be a potential research direction for MAT-MI.

Finally, after all the technical improvement, in vitro or in vivo animal experiments need to be further conducted to assess the MAT-MI method for high-resolution conductivity imaging. Different types of soft tissues, organs or the whole body may be explored in the future imaging studies. The most possible clinical application of MAT-MI is believed to be tumor
detection especially in cancer screening and detection. A recent work on magneto acoustic tomography with the aid of magnetic nanoparticles suggested such possibility; employing the magnetomotive force, magneto acoustic tomography in a MAT-MI compatible setup was demonstrated as a high-spatial-resolution approach to reconstruct the in vivo distributions of nanoparticles, and to further highlight tumor by a high contrast image (Mariappan et al 2016). This study sheds light on the possibility of improving magnetoacoustic imaging by harnessing the versatility of nanoparticles. Without using the contrast agents, a high-frequency MAT-MI technique has been developed lately (Yu et al 2016). This study has demonstrated the first in vivo tumor image by the MAT-MI, and the improved spatial resolution to detect not only the muscle-tumor interface but also the internal conductivity variations of the tumor. As reported, the MAT-MI has also been demonstrated to track the tumor growth on a tumor-bearing mouse model, which shows the feasibility of applying this imaging technique for early cancer detection (Yu et al 2016).

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