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MREIT with SENSE acceleration using dedicated RF coil design

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Abstract

We have adapted SENSE-accelerated imaging technique to Magnetic Resonance Electrical Impedance Tomography (MREIT) in order to acquire data faster or improve spatial resolution without increasing the data acquisition time. We also designed and optimized an 8-channel, dedicated phased array coil for this application. The RF coil was designed as two planar arrays so that the open sides could accommodate the MREIT electrodes. A quasi-intrinsic decoupling scheme was used to minimize couplings between coil elements and a restricted geometry optimization was applied to achieve high and uniform SNR within a given coil geometry and animal size. The impact of variability in the size of animals was also investigated in the optimization. This coil geometry provided approximately 3 times higher SNR at the center of a phantom when compared to a transmit/receive birdcage coil designed for MREIT of small animals. Therefore, the SNR was superior to the data acquired with volume coils even with three times SENSE acceleration. Both magnitude images and MREIT data were acquired from phantoms with and without SENSE acceleration and the results were compared. The phantom studies demonstrated that conductivity maps can be reconstructed from MREIT data acquired with 3 times SENSE acceleration and that no discernible artifacts were observed when accelerated data acquisition was compared to non-accelerated data.

Keywords

MREIT; SENSE imaging; MRI RF coil design

1. Introduction

Electrical Impedance Tomography (EIT) has a range of biological applications, such as detecting tumors (Surowiec et al 1988, Cherepenin et al 2001, da Silva et al 2000, Muftuler et al 2006a). This technique was first proposed in the early 1980s to reconstruct the conductivity distribution inside an object. In EIT, a current distribution is generated inside the object by current injection or induction and peripheral voltage measurements are acquired to find the internal conductivity distribution (Brown and Barber 1987). However, this method suffers from poor spatial resolution due to several factors. First of all, the limited number of voltage measurements taken only from the surface reduces the sensitivity to conductivity changes deep inside the object. Secondly, the currents do not follow a linear path inside the object, nor do they follow a single path, which has an adverse effect on the accuracy of the reconstruction.

Magnetic Resonance Electrical Impedance Tomography (MREIT) is an alternative technique for imaging conductivity (Woo *et al* 1994, Ider and Birgul 1998). It uses the magnetic flux density measurements acquired from MR phase images to reconstruct conductivity. A single

component of the magnetic flux density generated by applied currents can be measured with high spatial resolution using MRI. Unlike conventional EIT, the spatial resolution in the MREIT is uniform throughout the object. On the other hand, the conductivity reconstruction should be obtained from this single component of the magnetic flux density (hereafter called b_z), which is parallel to the main field of the MRI. Rotation of the object in the MRI is required to acquire all three components of the magnetic flux density. However, this is impractical and time consuming in most cases.

Once the b_z is obtained from the MRI phase maps, one can reconstruct the relative conductivity distribution inside the object. An additional surface voltage measurement is required to determine the absolute conductivity values (Kwon *et al* 2002, Birgul *et al* 2003a). Several methods have been proposed to reconstruct conductivity images from b_z maps, (Birgul *et al* 2003b, Woo *et al* 1994, Kwon *et al* 2002, Seo *et al* 2003).

A closely related technique called MR Current Density Imaging (MR-CDI) provides maps of the current density distribution inside an object instead of the conductivity. This technique can be utilized to identify the current paths of defibrillators (Joy *et al* 1989, Yoon *et al* 2003, DeMonte *et al* 2006). It can also potentially identify the current paths of Transcranial Magnetic Stimulation (TMS) with minimally invasive surface electrodes instead of the methodology used by Wagner *et al* (2004). Several researchers developed techniques to reconstruct current density images instead of conductivity images using magnetic flux density maps, but they require knowledge of all three components (Joy *et al* 1989, Yoon *et al* 2003, DeMonte *et al* 2006).

In all these MREIT and MR-CDI techniques, with the exception of (Hamamura *et al* 2008), a variant of the conventional spin-echo or gradient echo (Oh *et al* 2005) pulse sequence was utilized in acquiring the data, which leads to long imaging times. In order to shorten the data acquisition time (or increase the spatial resolution without increasing the data acquisition time), we adapted the Sensitivity Encoding (SENSE) method (Pruessmann *et al* 1999) to MREIT. This technique reduces total data acquisition time by undersampling the MRI data in the phase encode direction. The full-field image is reconstructed from the resultant aliased MR image using individual coil sensitivities. Application of SENSE imaging to MREIT requires a dedicated RF coil design to accommodate the MREIT electrodes and their support. Moreover, MREIT reconstruction is sensitive to noise in the phase maps and accelerated SENSE imaging produces images with a spatially nonuniform noise distribution. Therefore, it is essential to utilize a coil optimization in order to produce an image with SNR as uniform as possible, within the given geometrical limitations. For this purpose, an RF coil optimization technique that was previously reported by our group for head imaging was modified and applied to the new MREIT geometry to improve coil performance (Chen *et al* 2007, Muftuler *et al* 2006b). Coil optimization is important in order to ensure that the geometry factor of the coil will be minimized and the SNR of the SENSE accelerated MREIT data will still be superior compared to those acquired with conventional volume coils.

In the next section, we describe the design of a phased array coil optimized for MREIT applications on animals. In the following sections, the optimization method, performance evaluations with and without SENSE acceleration and phantom results are presented.

The coil geometry was optimized to achieve improved SNR and uniformity by using a method that was described in section 2.1.1. We also studied the variation of coil performance with different animal sizes and a design was constructed which would perform the best for a range of animal sizes. This is especially important because animal sizes vary substantially in a typical study, making it desirable to pick a coil design that performs optimally for a range of animal sizes. Data were acquired from conductivity phantoms prepared for this study and results were

presented in section 3. The SNR calculations for full-FOV and SENSE accelerated MREIT data were compared and noise calculations were presented for conductivity images. The image SNR performance was also compared to that of a conventional birdcage coil, which was currently used for MREIT experiments. The receive-only, 8-channel phased array coil designed for this study provided approximately three times higher SNR compared to the transmit/receive birdcage coil that we had built for MREIT studies.

2. Methods

2.1. RF coil design

The receive-only coil was designed as two planar arrays, each with four elements. The overall topology is illustrated in Fig.1. In such 8-element RF coils, there will be mutual coupling between 28 different coil pairs. Therefore, proper design topology to minimize coupling is of the utmost importance.

Coil overlapping is a simple technique to reduce mutual coupling between adjacent coils. However, overlapping reduces the effective surface area of the coil for signal detection, thus reducing efficiency. Moreover, overlapping coils have significantly large overlapping sensitivity profiles, leading to poor SENSE imaging performance. Therefore, a butterfly shape adjacent to rectangular loops was utilized to reduce coupling between these adjacent elements. The two wings of the butterfly coil generate magnetic flux density in opposite directions in the adjacent coil loop. This generates currents equal in amplitude but opposite in direction, thus cancelling each other out. As a result, coupling between co-planar pairs 1–2, 3–4, 5–6 and 7–8 as well as opposing pairs 1–5, 2–5, 3–7 and 4–8 will be minimal. On the other hand, geometrically similar pairs 2–3 and 6–7 were decoupled using hand-wound transformers. Minimizing the amount of mutual coupling between these decoupling transformer windings was achieved by adjusting the distance between the transformer windings or by changing the number of turns of the transformers. This was done heuristically until the coupling was below -18dB , as measured by an HP 4395A network analyzer. Distant coil elements already had negligible coupling due to their size and separation, so further decoupling was not needed. Although we have not seen reports of phased array coils with the geometry identical to the ones used here, similar architectures utilizing combinations of butterfly and loop shaped elements have been reported by others (Ohliger *et al* 2005).

2.1.1. Coil Optimization—The general approach in coil optimization is to describe the coil in terms of simple geometric shapes. These shapes can be parameterized, and the parameter values optimized in order to yield, for example, a uniform SNR or a highest mean SNR. The coil can be modeled as the spatial distribution of surface current density on a predefined former (Muftuler *et al* 2006b) or it can be modeled as a set of connected conductor segments (Chen *et al* 2007). Once the coil structure is defined and the magnetic flux density generated by that structure is calculated, then the SENSE imaging SNR can be formulated as a function of the RF coil geometry and a solution can be sought in the least squares sense to find the coil structure with the best performance within the restrictions imposed on the geometry. Performance can be defined in different ways; one can attempt to get the highest mean $\text{SNR}_{\text{sense}}$ inside a target volume or achieve a more uniform SNR. Various target volumes can be defined for the optimization.

A coil element such as the butterfly shape in Fig.2 is made up of straight wire segments combined to form a closed loop. The total magnetic field of the coil can be calculated as the superposition of the fields from each wire segment. Each wire segment can be defined in terms of the location of its vertices $\{r_i\}_\gamma$, where γ represents each coil element. The numbering of the vertices describing the coil was chosen in such a way that the positive current direction corresponds to ascending vertex numbers. Thus, the magnetic field of each element can be

defined by the locations of these vertices, and these locations can be used as the parameter set for optimization. In order to perform this optimization, one has to establish an analytic relationship between $\text{SNR}_{\text{sense}}$ and the locations of coil vertices. The SNR at pixel p in SENSE imaging is given by:

$$\text{SNR}_{\text{sense},p} = \text{SNR}_{\text{full},p} / (g_p \cdot \sqrt{R}) \quad (1)$$

SNR_{full} is the SNR when no SENSE acceleration is used, R is the SENSE acceleration factor and g_p is the geometry factor (g -factor) that is given by:

$$g_p = \sqrt{\left((\Sigma^H \cdot \Psi^{-1} \cdot \Sigma)^{-1} \right)_{p,p} (\Sigma^H \cdot \Psi^{-1} \cdot \Sigma)_{p,p}} \quad (2)$$

Σ is the coil sensitivity matrix (H indicates transposed complex conjugate) and Ψ is the receiver noise matrix that is approximated by summing the scalar product of the coil sensitivities over a number of points inside the volume of interest (VOI) (Pruessmann *et al* 1999, Weiger *et al* 2001). SNR_{full} was calculated using the root-sum-of-squares combined pixels as described in Roemer *et al* (1990), which was also adopted by DeZwart *et al* (2002).

In this calculation, it was assumed that the sample noise would dominate, and thus the noise components associated with coil conductor losses, matching circuits and preamplifier were ignored. A uniform conductivity distribution was assumed in the VOI for the calculation of the sample noise. As a result, an analytical relationship between the $\text{SNR}_{\text{sense}}$ and the coil magnetic flux density distributions was established and optimization was achieved using a least squares approach. For example, the reciprocal of the mean of the $\text{SNR}_{\text{sense},p}$ can be minimized, maximizing the overall SNR.

2.1.2. Optimum coil performance and variations in animal sizes—Since the coil was specifically designed for use in MREIT studies on small animals, a digital rat model shown in Fig. 3 was used to define the VOI for optimization.

Given that there can be a significant amount of variability in body size and weight in experimental animals, we investigated how variations in body size affected the overall coil performance. For this purpose, the digital rat model was scaled proportionally from 100% to 80% with 5% decrements and the coil optimization routine was run separately five times to obtain the coil design parameter sets for each animal size. Then, imaging simulations were run with each of the resulting RF coil arrays on these five different VOI sizes to observe which coil provided the best $\text{SNR}_{\text{sense}}$ performance for the specified range of animal sizes. All coil arrays were optimized for a SENSE acceleration factor of 3 and the same imaging parameters were assumed for comparison. Since the basic topology for the whole coil array shown in Fig. 1.a is required for efficient decoupling, only the coil dimensions (length and width of the elements) and gap sizes were allowed to vary for the optimization by moving the coil vertices within the plane. Restrictions were imposed to avoid coil overlaps.

2.1.3. Phased array coil implementation—In order to cover the whole volume of a typical rat body, the overall width and length of the top and bottom coil elements were chosen to be 68mm and 130mm, respectively (Fig.1.a). The elements were assembled onto two 3mm thick flat acrylic plates. 0.06mm thick copper tape (3M St. Paul, MN) was fixed onto the acrylic plates in 3mm wide strips to build the coil. Then each coil element was tuned to the 4T resonance frequency of 170MHz using distributed ceramic capacitances. Solenoid BALUNs were built in-house and used to isolate the cable shield from the coil ground. Passive detuning

during RF power transmit was achieved using crossed switching diodes and a parallel resonant circuit. The signals from the coil elements were amplified using eight surface mount monolithic amplifiers (model RAM-8, Mini-Circuits, Brooklyn NY).

2.1.4. Birdcage RF coil design for MREIT experiments—A conventional quadrature birdcage coil was also designed and built for our MREIT experiments. The SNR performance of the designed array coil was compared against this birdcage coil. Even though it is well known that array coils offer improved SNR compared to volume coils, this SNR drops with SENSE acceleration. Therefore, by quantifying the improvement in SNR, one can predict the amount of SENSE acceleration which yields an $\text{SNR}_{\text{sense}}$ that is still better or comparable to that of the volume coil. The birdcage coil was a 16-rung highpass type with 100mm diameter and 160mm length. The size of the coil was chosen to accommodate large size Sprague-Dawley rats with an animal holder to stabilize both the animal and electrodes in a fixed frame.

2.2. MREIT with SENSE acceleration

In MREIT, electrical currents are injected into an object, generating a volume current density inside the object. These internal currents generate magnetic fields, the z component of which encodes additional phase information in the MRI signal (here the direction of MRI main magnetic field is assumed to be the z direction):

$$S^{\pm}(\mathbf{U}) = \iint M(\rho) e^{j\theta(\rho)} e^{j\mathbf{U} \cdot \mathbf{r}} e^{\pm j\gamma b_z(\rho) T_{\text{pulse}}} dx dy \quad (3)$$

$S^{\pm}(\mathbf{U})$ represents the complex k-space data with injection of currents with $\pm$ polarities. The static phase shifts due to various sources such as static field inhomogeneity and other hardware related phase delays are all summarized in the phase term $\theta(\rho)$. Since this static phase term is unknown, the data has to be collected twice with the polarity of electric current reversed. When the phase of these two MREIT images are subtracted, the static phase terms will be eliminated, leaving only the phase term introduced by $b_z(\rho)$. In equation 3, the longitudinal (T1) and transverse relaxations (T2) were ignored. In fact, they can be considered as constant weighting factors due to the short acquisition window, and thus can be viewed as a scaling factor of $M(\rho)$, which is the bulk magnetization of protons at location (ρ) . γ is the gyromagnetic ratio, and $\mathbf{U} = [u \ v]$ is the vector of spatial frequencies given by $u = \gamma^* G_x^* T_{Gx}$ and $v = \gamma^* G_y^* T_{Gy}$, where T_{Gx} and T_{Gy} are the durations of the gradient pulses G_x and G_y , respectively. As a result, the total phase accumulated at a pixel at location (ρ) due to the magnetic flux density generated by injected currents is given by:

$$\varphi(\rho) = 2 \cdot \gamma \cdot b_z(\rho) \cdot T_{\text{pulse}} \quad (4)$$

where $b_z(\rho)$ is the magnitude of the z-component of the magnetic flux density at location (ρ) generated by electrical currents inside the object and T_{pulse} is the duration of one lobe of the bipolar current pulse. The phase term in equation 4 can be easily measured from the MREIT images and the z-component of the magnetic flux density distribution can be calculated.

A schematic of the MREIT pulse sequence is shown in Fig.4.

2.3. Reconstruction of MREIT data using SENSE

In order to obtain the $b_z(\rho)$ in equation 4, full field images have to be reconstructed first from the aliased SENSE accelerated images. In order to preserve the phase information for each

pixel, unfolding was done in the complex domain. To unfold the aliased images, we followed the basic procedure described by Pruessmann *et al* (1999).

Once the complex pixel values for the full image are reconstructed from the SENSE accelerated image, the magnetic flux density generated by the internal currents is calculated from the phase value at each pixel using equation (4).

Various methods have been proposed to obtain conductivity from $b_z(\rho)$ measurements, such as the sensitivity matrix method (SMM) (IDER *et al* 1998, Birgul *et al* 2003b), $\nabla^2 B_z$ algorithms (Seo *et al* 2003, Ider *et al* 2004), and ∇B_z algorithms (Park *et al* 2004a, Park *et al* 2004b). We utilized the SMM method for MREIT reconstructions. In this technique, a discretized mesh of first order triangular elements was defined for the object domain in order to utilize the finite element method (FEM) for numerical calculations. In order to construct conductivity maps from $b_z(\rho)$ measurements, a first order linear approximation was assumed between conductivity perturbations and z -component magnetic flux density perturbations:

$$\Delta \mathbf{B} = \mathbf{S} \cdot \Delta \sigma \quad (5)$$

Here $\mathbf{S}$ is called the Sensitivity Matrix. If there are m measurement points and n conductivity elements, $\Delta \mathbf{B}$ is an $m \times 1$ vector, $\Delta \sigma$ is an $n \times 1$ vector, and $\mathbf{S}$ is an $m \times n$ matrix.

In general, $\mathbf{S}$ is an ill-conditioned matrix, thus convergence may not be achieved by solving equation (4) using an ordinary least squares approach to calculate σ_{final} . Therefore, the equation was solved using the conjugate gradient method with Tikhonov regularization, where σ_{final} was found by:

$$\sigma_{final}^{min} \left\{ \|\Delta \mathbf{B} - \mathbf{S} \Delta \sigma\|^2 + \lambda \|\Delta \sigma\|^2 \right\} \quad (6)$$

where λ is a regularization parameter (Golub 1999). The regularization parameter λ was chosen such that the calculated z -component magnetic flux density generated by the reconstructed conductivity distribution was closest to the MRI-measured magnetic flux density. The details of the forward problem and SMM were published in (IDER and Birgul 1998 and Birgul *et al* 2003) and have been omitted here.

2.4. Phantom Preparation

A phantom with a non-uniform conductivity distribution was prepared to test the SENSE accelerated MREIT (Fig. 5.a). The phantom was built inside a hollow acrylic cylinder with a 44mm inner diameter. The phantom consisted of a 10 mm thick layer of agarose gel, which was made by mixing deionized water with 4 mM CuSO₄ and 2% agarose (g/100 ml water). Within this disc, a smaller cylindrical region of 12 mm diameter and 1 cm thickness was filled with 4 mM CuSO₄, 2% agarose (g/100 ml water) and 0.24% NaCl. This resulted in a conductivity ratio of $\sigma_I/\sigma_{II}=5.6$. Four 6.35 mm wide copper electrodes were placed equidistant along the inner wall of the acrylic disc and used to inject currents into the interior region. Currents were delivered to the electrodes with thin copper wires that were placed parallel to the main static field of the MRI so that the magnetic fields generated by these currents had a negligible effect on the MREIT measurements. The phantom was placed in the scanner such that the axes of the cylindrical phantom and the MRI magnet were parallel.

2.5. Data Acquisition

Data were collected using a 4T whole body (human size) MRI magnet (Magnex Scientific Inc., UK), which was interfaced to an ISOL ISP-400 spectrometer (ISOL Technology Inc., Korea).

The system is equipped with 8 broadband RF transmit and receive channels and a whole body gradient coil set (Tesla Eng. UK). The system also includes a 13 channel room-temperature high-order shim system with MXA-13-4 shim power supply (Resonance Research, Billerica, MA) to minimize field inhomogeneities.

2.5.1. Image SNR comparisons—The SNR performances of the designed phased array coil and the conventional birdcage coil were compared through a simple imaging experiment. Images were taken in the axial orientation using a gradient echo pulse sequence with TR=350ms, TE=12ms, flip angle=30 degrees, 2.0mm slice thickness, 1.0mm gap and 100mm FOV. The acquisition matrix was 128×128. Two hollow acrylic cylinders with 38mm inner diameter were filled with 4mM CuSO₄ solution and used for SNR tests. Two circular regions of interest (ROI) with 10mm diameter were placed at the center of each phantom and mean pixel intensity was calculated. To calculate noise, the standard deviation was measured from the background and multiplied by 1.53 to account for the rectified noise in the magnitude images (Sadleir *et al*, 2005). Then SNR is evaluated as the mean divided by the noise.

2.5.2. MREIT experiment—An MREIT experiment was conducted using the two-compartment cylindrical phantom described in section 2.4. First, an MR image was collected without SENSE acceleration and without any current injection to acquire the sensitivity information of the RF coil array using a spin echo pulse sequence with parameters TR/TE=500ms/50ms, slice thickness=5mm, FOV=80mm, matrix size=64×64. The combined magnitude image from all coils is shown in Fig 5.b. Then, two MREIT data sets were acquired, one with no SENSE acceleration and another using 2.7 times SENSE acceleration. For each case, data were acquired for 2 different current injection profiles. A 1mA bipolar current pulse ($T_{\text{pulse}}=15\text{ms}$) was injected between diagonal electrode pairs 1–3 for the first run and 2–4 for the second. Each data set was acquired twice with opposite polarities. The two phase maps obtained with opposite polarities of current injected through a particular pair of electrodes were subtracted to remove all phase from other static sources and obtain the phase shift due to the magnetic field generated by the injected currents. All images were taken in the axial direction using the spin echo based MREIT pulse sequence (Fig. 4), with left-right phase encoding direction. A single 5mm thick slice was acquired with TR=500ms and TE=50ms. The acquisition matrix was 64×64 for the full-FOV sensitivity measurement and the FOV was 80×80mm. For the SENSE measurements the acquisition matrix was 64×24 and the FOV was 80×30mm, so the acquisition speed was increased by approximately 2.7 times for each MREIT phase measurement with SENSE. With these sequence parameters, the duration of one full-FOV MREIT acquisition was 32s. Therefore, with two polarities of the injected current and two different current injection profiles, it took 128s to acquire one set of MREIT data. On the other hand, the total data acquisition time with 2.7 times SENSE acceleration was only about 47 seconds.

2.5.3. Noise Calculations in MREIT data—Noise in b_z images was calculated following the noise analysis technique reported by Sadleir *et al* (2005). In that article, two alternative methods were proposed to estimate the noise in the b_z images. In our calculation, the first approach was used in which the noise was estimated using:

$$sd(b_z(x, y)) = \frac{1}{\sqrt{2} \cdot \gamma \cdot T_c \cdot \Gamma_M} \quad (7)$$

In Equation (7), γ is the gyromagnetic ratio, T_c is the total duration of the injected current ($2 \times T_{\text{pulse}}$) and Γ_M is the SNR of the MR magnitude image. In order to estimate Γ_M , two ROIs were selected in the MR magnitude images from both SENSE accelerated and non-accelerated

data sets. In an MR image that is acquired using phased array coils, the spatial distribution of SNR is not uniform; therefore two ROIs were used to assess the spatial variation of SNR. The first ROI was selected at the center of the phantom and the second one was placed towards the side of the phantom at equal distances from both coil planes. This second ROI was placed in a region where there was aliasing due to SENSE acceleration. This region is expected to have lower SNR compared to the center of the object, where there were not any aliased pixels. Care was taken to place the ROIs in the same locations in both accelerated and non-accelerated images. The mean signal intensities and the standard deviations were calculated from these ROIs, as suggested by Sadleir *et al* (2005). A region outside the object was not selected for noise calculations because one can only estimate the coil sensitivity information outside the object using extrapolation. Therefore, the background noise in a SENSE-reconstructed image would not be an accurate representation of the actual noise.

3. Results

3.1. Image SNR: SENSE Array versus Birdcage

Images were acquired using the gradient echo pulse sequence described in section 2.6 and SNR values were calculated as described in 2.5.1. The birdcage and the coil array had SNRs of 90 and 264, respectively. The average SNR in the outer region of the phantom images was around 503 for the coil array since the sensitivity is higher in the vicinity of the coil elements. Therefore, significant SNR improvement in MREIT studies was achieved with the new 8-channel array coil.

3.2. Coil performance based on target volume size used in optimization

The coil optimization algorithm was run on five different animal sizes and, as a result, five slightly different coil designs were obtained (see Fig.1 and table 1 for the coil geometry and dimensions). Five separate SENSE accelerated imaging simulations were done on each coil design using five different animal sizes in order to evaluate the performance of each coil for different animals. Figure 6 summarizes the results of these simulations. Horizontal axis is the animal body size normalized to the largest size considered in simulations. Vertical axis is the lowest SNR_{sense} measured in the coil. Since SNR is spatially nonuniform in a phased array coil, the lowest SNR in the simulated images were investigated.

Each of the curves represents the lowest SNR measured with different animal sizes inside a coil that is designed for a specific animal size. For example, the **100% coil** line shows the performance of the coil array designed with the largest animal considered for this study (we will call this the **100% coil**). As seen in figure 6, the coil array designed for the largest animal showed the lowest SNR of 79 in an image of the animal with 80% body size, and lowest SNR of 74 for a 100% animal. On the other hand, the coil designed for 80% animal size (80% coil) shows much lower SNR of 64 for a 100% animal, while its performance for an 80% animal is similar to the 100% coil.

On the other hand, all coils had similar mean and maximum *g-factors* for three times SENSE acceleration for all body sizes considered.

3.3. MREIT experiment

The phase difference images obtained with 2.7 times SENSE acceleration are illustrated in Fig. 7. Figures 7.a and 7.b are the two phase maps obtained with current injections from electrode pairs 1–3 and 2–4, respectively. The conductivity images were calculated using equation (6) and the results are illustrated in figure 8–figure 9. The MREIT image reconstructed from the data with 2.7 times SENSE acceleration is shown in Fig.8 and the one with no SENSE acceleration is shown in Fig.9.

Three ROIs were defined inside these conductivity images reconstructed from the full FOV and SENSE accelerated images. The first ROI encircled the center 5mm diameter portion of the high conductivity region I of Fig.5.a. The second ROI encircled the whole high conductivity region and the last ROI encompassed the background segment of the phantom (region II). Average values and standard deviations were calculated within these ROIs and are summarized in table 2.

3.3.1. Noise Calculations in MREIT b_z images—The noise measurements in b_z images were carried out using the technique described in section 2.5.3. The SNR measurements inside the two ROIs in the magnitude images with and without SENSE acceleration were summarized in table 3. These SNR measurements were used in equation (7) to calculate the noise in b_z images. Table 4 lists the calculated noise in b_z for full-FOV and SENSE accelerated MREIT. Typical noise was around 0.9nT in b_z that was obtained from the full-FOV data, while it increased to about 1.5nT ~ 1.7nT with the 2.7 times SENSE acceleration. As a reference, the maximum and minimum b_z values in the data were 20.1nT and -18.3nT, respectively.

4. Discussion and conclusion

Through this study, it has been demonstrated that the SENSE accelerated imaging technique can be applied to MREIT to shorten the data acquisition time (or increase the spatial resolution) of MREIT experiments. But, it should be emphasized that it is important to optimize the RF coil design to achieve high SNR with uniform distribution for the object or animal that will be investigated. In MREIT, the quality of the reconstructed conductivity maps is sensitive to the noise in the phase maps obtained from the MR images. Since the spatial distribution of noise is nonuniform in SENSE imaging, such noise nonuniformity might be reflected as strong systematic artifacts in the conductivity images. The amount of noise nonuniformity in SENSE accelerated imaging is influenced by the acceleration factor and the coil sensitivity profiles. Therefore, an optimal coil configuration is essential to minimize such artifacts. For that purpose, we designed and implemented an 8-element RF receive coil array. The coil was designed to accommodate mice and rats of various sizes together with associated MREIT electrode holders. The coil sensitivity was optimized, within given geometrical constraints, using an inverse method that we developed for optimizing SENSE array coils for various applications (Chen *et al* 2007, Muftuler *et al* 2006b). Since the optimization of SNR_{sense} is realized for a given target volume, the variation of SNR_{sense} with different body sizes was also investigated. It was observed that if the coil is optimized for the largest body size anticipated, it gives the best performance for the range of body sizes considered for the experiments. On the other hand, the maximum *g-factor* of coils designed for various body sizes did not show such a dramatic change across different body sizes (at 3 times SENSE acceleration).

In this study, quasi-static field approximations were used for the field calculations since the dimensions of the coil elements and the animals are small compared to the wavelength of the 4T resonance frequency (170MHz). This provided an optimization algorithm that is computationally less expensive, while providing sufficient accuracy in the calculations of field distributions. For more accurate calculations, especially for larger RF coils, one should utilize the full-field equations implemented by numerical analysis techniques, such as FDTD.

SNR and noise calculations were carried out on the images acquired with the 8-channel coil array as well as the birdcage coil that was currently in use for our MREIT studies. First, the SNR values in full-FOV magnitude images were measured for the designed SENSE coil and results were compared to those of the birdcage coil. It is already known that phased array coils generally provide higher SNR compared to volume coils; but we have demonstrated that even with SENSE acceleration, the SNR would be superior to the volume coils. This way the experimenter could afford to collect data with fewer averages, which will also reduce the total

imaging time. As shown in table 2, the SNR of the magnitude images dropped by 1.73 times at the center ROI and 1.85 times inside the second ROI when 2.7 times SENSE acceleration was used. If the *g-factor* ideally had a value of 1 everywhere, one would expect a SNR drop of 1.64 (see equation 1). However, in practice, the *g-factor* is generally closer to 1 at locations where there are no aliased pixels and it is higher where there are mostly aliased pixels. Therefore, spatially varying SNR was expected. Since the noise in b_z scales inversely proportionally as given in equation (7), it increased in the SENSE accelerated images. However, the noise was still more than an order of magnitude smaller than the peaks in the b_z maps.

We have used a relatively simple approach to calculate and compare SNR values. However, those measurements are not highly accurate due to spatially varying noise values in full-FOV and SENSE accelerated images of phased array coils. The approach used here provides an estimate of the average SNR within the chosen ROI. However, one can get more accurate SNR measurements using the technique proposed by Kellman and McVeigh (2005).

Since the conductivity maps are reconstructed from only one component of the magnetic field generated by internal currents and the relationship between this magnetic field component and the conductivity distribution is nonlinear, any systematic artifacts in SENSE based reconstruction might lead to major artifacts in MREIT based conductivity images. However, we did not encounter any systematic artifacts in the reconstructed conductivity images. The SNR of the conductivity image within the high conductivity region was 53 for the case where no SENSE acceleration was used. Similarly, the SNR in the same region for SENSE accelerated MREIT was 46. The conductivity ratio of region I to region III [$\text{mean}(\text{ROI } 1)/\text{mean}(\text{ROI } 3)$] was 5.6 in the no-acceleration image. This is exactly the conductivity ratio calculated based on the ion concentrations described in section 2.4. On the other hand, the ratio of region I to region II was calculated to be 5.46 in the SENSE accelerated image, which is only 2.5% lower than the actual value. We did not use the full size of the high conductivity region (ROI 2) for this calculation because of ion diffusion, which causes a drooping edge spread function (Hamamura *et al*, 2006). Even though the mean values given in table 2 for the accelerated image appear to be lower compared to the no-acceleration images, it should be noted that the conductivity images in MREIT are arbitrarily scaled and the measurements are relative conductivity values. Absolute conductivity values can be obtained only if external voltage measurements are taken simultaneously (Kwon *et al* 2002, Birgul *et al* 2003a). The arbitrary scaling might be affected by several factors such as the (automated) choice of the regularization parameter or small drifts in the MRI system.

When the two data are analyzed, the conductivity uniformity appears to be better in SENSE accelerated imaging. This is possibly in part due to minor fluctuations in the system. Slight variations in gradient performance may lead to fluctuations in signal phase, which may not be detectable in the MR magnitude images but can have a stronger effect on the MREIT images. Stability of the system would be generally better during shorter imaging times, thus may lead to reduced artifacts in MREIT if SENSE acceleration is used. This issue will be investigated further.

The SENSE imaging technique can be utilized for various purposes in MREIT experiments. For example, it can be used to acquire data in a much shorter time. In a typical MREIT experiment four images are acquired sequentially injecting current from two different pairs of electrodes and two opposite polarities of current. SENSE acceleration would significantly shorten this data acquisition. As demonstrated here, the total MREIT data acquisition time was reduced to 47s from 128 s. Moreover, the SNR is still better than a birdcage volume coil even with 2.7 times SENSE acceleration. Therefore, one can also reduce the number of averages and shorten the image time further without SNR penalty, when compared to studies conducted

with volume coils. With shortened data acquisition times, it may also be possible to monitor dynamic changes in conductivity using MREIT. On the other hand, it is also possible to collect higher spatial resolution MREIT data within a given time using the SENSE technique.

In the presented work, we used a moderate SENSE acceleration of 2.7 to ensure that the SNR is not significantly degraded. It is theoretically possible to go to higher SENSE accelerations using the 8-channel coil array. When the noise variance and maximum values in the b_z images are considered (section 3.3.1), it can be seen that one can go to five times acceleration rates and still keep the SNR in b_z images reasonably high. But it should also be kept in mind that the phantom was not thick in the z-direction, thus the electrical currents were mostly confined to the x–y plane. In animals, there would be a three dimensional distribution of currents and the maximum values in b_z images would drop. Another important factor in deciding the acceleration rate is the coil design. The coil optimization is performed for a defined acceleration rate; therefore it performs optimally at acceleration rates near that value.

As mentioned earlier, there is also a blurring effect seen on the edges between the two compartments. This is mostly due to the diffusion of ions from the higher concentration compartment to the lower concentration compartment. Therefore, the smooth edges do not directly reflect the line spread function of the imaging technique. We have shown this diffusion effect by acquiring time-lapse MREIT images of NaCl diffusion from one compartment to the other over a 12 hour period (Hamamura *et al* 2006).

Using the array coil developed for this study, we have collected MREIT data from a conductivity phantom using 2.7 times SENSE acceleration. A phantom with known geometry and conductivity distribution was utilized in order to assess the results and make comparisons. SENSE reconstruction of the aliased images was performed in the complex image domain to preserve phase information and full-field phase images were reconstructed from the data. The reconstructed conductivity images did not show any systematic artifacts due to SENSE acceleration. Interestingly, the drop in SNR of the conductivity images with SENSE acceleration was less than the drop in raw SNR in magnitude images. We anticipate that the iterative reconstruction based on sensitivity matrix equation that employed Tikhonov regularization is tolerant to increased noise, thus the drop in SNR in conductivity images is less than the drop in MR image SNR. Noise propagation in the iterative SMM technique with regularization will be investigated in future studies to verify this observation.

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Fig.1.

a. Basic structure of the two-plane, 8-element RF coil array. The values for the dimensions of the coil (L , w , g) and the distance D between the top and bottom elements were given in table 1 in section 3.2 for two different coils optimized for different animal sizes.

Fig.2.
Definition of coil geometry in terms of the vertices r_1 – r_8

Fig.3. Digital rat phantom used in VOI definition (left); the VOI that was used in calculations (right).

Fig.4. Schematic of the MREIT pulse sequence. Electrical current pulses of opposite polarity are applied before and after the application of the 180° RF pulse and the MRI data is acquired at the end of the second current pulse.

Fig. 5.

(a) The MREIT phantom with non-uniform conductivity distribution. Region I has higher conductivity than the background region II by approximately 5.6 to 1. The numbers on the circumference indicate the location of electrodes. (b) MR magnitude image of the phantom with no SENSE acceleration (custom phased array coil was used). ROIs used in the calculations of the signal mean and standard deviation of table 2 were shown on the magnitude image

Fig. 6.

Lowest SNR measured within the VOI for five array coil designs. Each of the lines represents the performance of one coil for different animal sizes. The 100% coil line represent the coil which is designed for the largest animal body suggested for this study, while the 80% coil represents the coil which is designed for an animal that has a body size, which is 80% of the largest. It is seen that the 80% coil shows poor performance for a body size of 100%, while 100% coil shows the best overall performance for the range of body sizes considered.

Fig. 7.
The phase difference images obtained with 2.7 times SENSE acceleration and 1mA current injection between electrodes 1–3 (a) and 2–4 (b)

Fig.8.

Conductivity image reconstructed from the MREIT data acquired with 2.7 times SENSE acceleration. A cross sectional profile is shown on the right. The theoretical profile is also shown with dashed gray line. The location of the profile is shown by the black dashed line on the conductivity image.

Fig.9. Conductivity image reconstructed from the data acquired without SENSE acceleration. A cross sectional profile is shown on the right. The theoretical profile is also shown with dashed gray line. The location of the profile is shown by the black dashed line on the conductivity image.

Table 1

Dimensions of the phased array coil optimized for the largest and smallest animal sizes considered for this study. (units are in cm)

	L	w	g	D
100% coil	9.0	2.48	0.28	5.86
80% coil	9.42	2.47	0.27	5.50

Table 2

Mean and standard deviation measurements inside three ROIs drawn in the conductivity images. ROI 1 is a 5mm diameter region inside the high conductivity compartment, ROI 2 encircles the whole high conductivity compartment and ROI 3 is the low conductivity background compartment. The location of these ROIs are shown in Fig.5.

	Full-FOV MREIT image		SENSE accelerated MREIT image	
	Mean	Std. dev.	Mean	Std. dev.
ROI 1	465	8.8	388	8.44
ROI 2	397	46.5	332	40.9
ROI 3	83	68	71	27.9

Table 3

SNR measurements in the images of the MREIT phantom of Fig.5.

SNR of magnitude image		
	Full-FOV MREIT image	SENSE accelerated MREIT image
ROI 1	99.6	57.5
ROI 2	93.7	50.7

Table 4

Noise in b_z , which is calculated using equation (7) and SNR values in table 3.

	Noise in b_z (in Tesla)	
	Full-FOV MREIT image	SENSE accelerated MREIT image
ROI 1	8.8×10^{-10}	15.3×10^{-10}
ROI 2	9.4×10^{-10}	17.4×10^{-10}