SENSE accelerated MREIT with optimized RF coil design

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Introduction:

SENSE imaging technique is adapted to MR Electrical Impedance Tomography (MREIT) to shorten acquisition times. A phased array coil was designed and optimized for this application and MREIT data was reconstructed from the SENSE accelerated images. A quasi-intrinsic decoupling scheme was used to minimize couplings between coil elements and the geometry was optimized to achieve high and uniform SNR. Data was acquired with and without SENSE acceleration and the results were compared. Phantom studies demonstrated that conductivity maps could be reconstructed from MREIT data acquired with 2.7 times SENSE acceleration with no discernible artifacts.

Methods:

The receive-only coil was designed as two planar arrays so that the open sides could accommodate the MREIT electrodes (Fig.1) (68mm×130mm). The topology shown was chosen for optimal decoupling because coil overlapping reduces efficiency and overlapping sensitivity profiles lead to poor SENSE imaging performance. In this design, the two wings of the butterfly coil generate magnetic flux density in opposite directions in the adjacent coil loop, thus cancelling mutual coupling. That way, coupling between co-planar pairs 1-2, 3-4, 5-6 and 7-8 as well as opposing pairs 1-5, 2-6, 3-7 and 4-8 will be minimal. Geometrically similar pairs 2-3 and 6-7 were decoupled using transformers. To optimize performance, SENSE imaging SNR must be formulated as a function of the RF coil geometry [1,2]. The total magnetic field of the coil can be calculated as the superposition of the fields from each wire segment. Hence, the position of each wire segment or its vertices can be used as the parameter set for optimization. SNR of pixel ρ in a SENSE accelerated image is given by: $SNR_{sense,\rho}=SNR_{full,\rho} /(g_\rho*\sqrt{R})$. Here both the full-image SNR_{full} and g_ρ are functions of the field distribution of coils, so an analytic expression was derived for SNR_{sense} as a function of coil vertices. Using a least squares approach, the position of vertices that maximized average SNR_{sense} within a target volume (VOI) was calculated. Since the basic topology of Fig.1 must be preserved, only coil dimensions and gap sizes were optimized. Since the coil is optimized for a target VOI, we have also investigated how body size affected overall coil performance. For that purpose, the VOI was scaled from 100% to 80% with 5% decrements and the coil optimization routine was run separately for each body size and 5 different coil geometries were obtained. Then, imaging simulations were run with each of the resulting RF coil arrays on these five different VOI sizes to observe which coil would provide the best SNR_{sense} performance for a variety of animal sizes. All coil arrays were optimized for a SENSE=3.

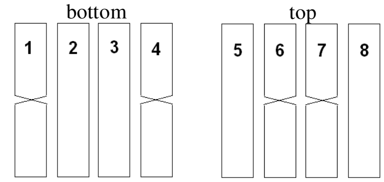

Fig.1. Basic structure of the two-plane, 8-element RF coil array.

In MREIT data, additional phase accumulates at a pixel at location ρ due to the z-component of the magnetic flux density, $b_z(\rho)$, which is generated by injected currents. This is given by: $\varphi(\rho) = 4\cdot\gamma\cdot N\cdot b_z(\rho) / \omega$ (N: # cycles of injected current, ω: angular frequency). To obtain the $b_z(\rho)$, first the full field images were reconstructed from the aliased SENSE accelerated images. In order to preserve the phase information for each pixel, unfolding was done in the complex domain. For three times SENSE acceleration, the maximum number of overlapping pixels would be 3, which results in 6 unknowns (real and imaginary parts); therefore one can obtain a unique solution with an 8-channel RF coil array [3]. Once the $b_z(\rho)$ was obtained, the conductivity image is reconstructed using the Sensitivity Matrix Method [4]. Data were collected using a 4T MRI magnet (Magnex Scientific Inc., UK), which was interfaced with an ISOL ISP-400 spectrometer (ISOL Technology Inc., Korea). MREIT data acquisition was conducted using a phantom that was built inside an acrylic cylinder with 44mm diameter. The phantom consisted of a 10 mm thick layer of agarose gel, which was made by mixing deionized water with 4 mM $CuSO_4$ 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 $CuSO_4$, 2% agarose (g/100 ml water) and 0.24% NaCl. This resulted in a conductivity ratio of σ_I/σ_{II}=5.6. Four copper electrodes were placed equidistant along the inner wall of the acrylic disc to inject currents. Two MREIT data sets were acquired with 1mA currents using either no acceleration or with SENSE=2.7. Details of the MREIT pulse sequence were given in [5].

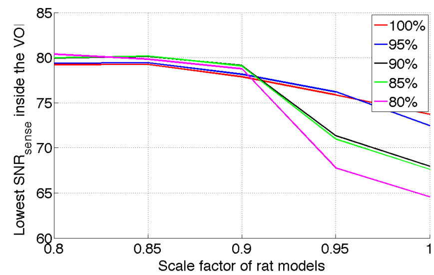

Fig. 2. SNR values for five coil designs. Each of the lines represents the performance of one coil for different animal sizes. e.g. 100% line represent the coil which is designed for the largest animal body suggested for this study.

Results and Discussion:

First, the SNR improvement over the conventional birdcage coil used for MREIT was investigated. Axial images were acquired using a gradient echo sequence (TR/TE=350ms/12ms, flip angle=30, slice thick:2mm, FOV=100mm, matrix:128x128). A phantom filled with 4mM $CuSO_4$ solution was imaged. The average SNRs within 10mm diameter circular region of interest (ROI) at the center of the phantom images were 90 and 264 for birdcage and phased array coils, 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 coil elements. Therefore significant SNR improvement was achieved. Fig.2 summarizes the results of how VOI size used in optimization affects overall performance for different animal sizes. Each one of the lines represents the SNR values for one coil, which is designed for a specific VOI size. In that figure, each curve shows the SNR performance of a specific coil for various animal sizes. It can be seen that the coil designed for the largest animal provides the best overall performance for various animal sizes. Finally, conductivity images with and without SENSE acceleration were investigated. Three regions of interest (ROI) were defined inside these conductivity images reconstructed from full FOV and SENSE accelerated images. The first ROI encircled the center 5mm diameter portion of the high conductivity region. The second ROI encircled the whole high conductivity region and the last ROI encompassed the background segment of the phantom. The mean and standard deviation were calculated within these ROI and summarized in table 1. Note that the conductivity ratios ROI1/ROI3=σ_I/σ_{II} are very close to the actual value in both cases.

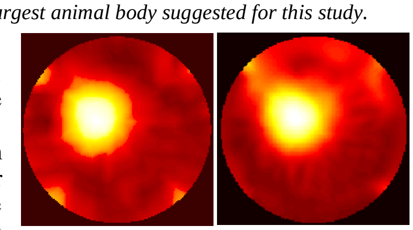

Fig.3. Conductivity images reconstructed from SENSE=2.7 data (left) and no acceleration data (right).

	Full-FOV		SENSE -MREIT	
	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 1. Mean and standard deviation inside the three ROIs drawn in the conductivity images.

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References: [1] Muftuler L.T., *et al* Physics in Medicine and Biology 51: 6457-6469, 2006; [2] Chen G., *et al* Journal of Magnetic Resonance 186: 273-281, 2007; [3] Pruessmann *et al* Magn. Reson. Med., 42: 952-962, 1999; [4] Ider YZ, Birgul O. Elektrik;6:215-225, 1998. [5] Muftuler L.T. *et al*, TCRT 5: 381-387, 2006.