

SENSE-Optimization of a Transceive Surface Coil Array for MRI at 4T

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Introduction

Typical fast parallel imaging (PI) techniques employ the use of radio frequency (rf) coils to transmit and receive the NMR signal. The signal intensity of each receiver is used to spatially encode image space (for SENSE [1]) for image reconstruction. The surface coil array has been shown to be an excellent choice of PI receiver due the two-fold benefit of providing the ability to spatially encode independent field profiles (required for PI) and it's high inherent signal to noise ratio (SNR)[2]. Typical receive-only SENSE configurations employ the use of two separate rf coils for transmitting and receiving the NMR signal which requires additional active decoupling between the two coils as well as decreasing the amount of available space within the bore of the magnet. In the current study we SENSE optimize a single surface coil array for both transmitting and receiving (transceive). The current study does not take into account other potential benefits of using SENSE for time series data studies, such as, the use of multiple navigators with multiple receivers, reduction of T_2^* blurring, and reducing macroscopic susceptibility artefacts.

Methods

A finite difference time domain (FDTD) method [3] algorithm is employed to solve Maxwell's electromagnetic wave equations in both time and space (3 mm isotropic spatial resolution) to simulate the performance of the transceive surface coil array. The transceive surface coil array (as shown in Fig. 1) consists of 8 surface coil elements that conform to a cylinder (27.9 cm diameter, 21.6 cm length). Each coil is made of copper with a width of the in-plane resolution (3 mm) and 8 distributed capacitors of equal value to tune the coil and minimize electric field loss. The coil separation is characterized by a variable [gap] defined as the ratio of the subtended angle of the gap to coil size [4]. In this study, we investigate 4 values of gap size [0.25, 0.5, 1.0, and 2.0] for SENSE optimization of at 4T. All variations of the surface coil array in this study are tuned to 170.3 MHz in simulation space using an impulse response method. In the case of the surface coil array in transmit mode, each capacitor is driven with an ideal current source to reduce the effects of magnetic coupling that would normally be eliminated by decoupling networks. A centered spherical (17.78 cm diameter) phantom is used to mimic the human head. The averaged brain tissue properties used were approximated from the frequency dependencies of the averaged conductivity and relative permittivity of brain tissue (Figure 6a [Yang]) at 170.3 MHz ($\epsilon_r = 86$ and $\sigma = 0.45$ S/m respectively). The FDTD simulations calculate the em rf fields of each transceive coil array to determine the power deposition (from electric field (E) loss), and signal intensity through the magnetic fields transmit (driven in a circularly polarized orientation) and receive profiles (β^+ and β^- respectively). This signal intensity is then used to calculate the geometry factor for accelerated Cartesian SENSE. Since both g and signal intensity are spatially varying, mean and maximum values are calculated for optimization.

A metric of performance E (Eqn. 1) is chosen to optimize the surface coil array for different gap sizes and acceleration rates as a function of the physiological-to-intrinsic noise ratio (PINR). In the simplest case of zero PINR, E simplifies to SNR/g, which is a conventional metric. The ratio of temporal noise for full and SENSE imaging ($\sigma_{i,full}/\sigma_{i,SENSE}$) can be expressed in Eqn. 1 based on [4]. The SNR is calculated by assuming a small tip angle regime where signal intensity can be expressed as a linearly with both the transmit and receive magnetic fields.

$$E = \left[\frac{\sigma_{i,full}}{\sigma_{i,SENSE}} \right] \cdot [SNR] \cdot \sqrt{R} = \left[\left(1 + (g^2 R - 1) \cdot \left(\frac{\sigma_{i,full}^2}{\sigma_{i,full}^2 + \sigma_{ph}^2} \right) \right)^{1/2} \right] \cdot \left[\frac{|\beta^+||\beta^-|}{\sqrt{\frac{1}{2} \int_v \sigma |E_{rms}|^2 dv}} \right] \cdot \sqrt{R} \quad [1]$$

Results

The calculations of E for the 3 separate cases of SNR (mean, center, and maximum) yielded similar results. For simplification, only the E calculated for mean SNR and mean g-factor are shown in Fig 2. The mean PINR within the human head was measured to be 1.3. Based on the measured PINR, an acceleration rate of 2 with a gap size of 0.25 is optimal (as read from Fig. 2) for MRI with this surface coil array at 4 T provided the maximum g-factor is not within the VOI.

Conclusions and Discussion

The benefits that parallel imaging promises to yield are directly related to the ratio of physiological to intrinsic noise contributions of the acquired images. For head imaging at 4 T, aforementioned design is optimum, at least for 8 channels. Based on our results, the optimum acceleration rate and coil gap size should increase for applications where physiological noise may be excessive (such as cardiac imaging)

An unexpected result is that the gap size (varying SNR and g-factor) does not play as critical a role in assessing optimum performance compared to the acceleration rate (R) and physiological to intrinsic noise ratio (as can be clearly seen in Fig 2). We have shown that the ratio of physiological to intrinsic noise is the predominant variable when choosing an optimum acceleration factor for PI. Emphasis must be placed on measuring the ratio of physiological to intrinsic noise for specific applications when choosing an acceleration factor that maximizes performance. Conversely, the rf surface coil array gap size appears to play a minor role in optimization (for volume imaging with a large FOV), suggesting that a single rf surface coil array would be versatile for implementation in multiple applications without a significant impact on performance.

References

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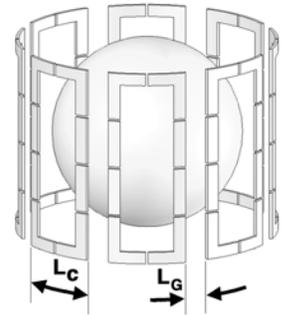


Fig. 1: The transceive surface coil array is shown for gap size of 0.25. A spherical phantom with head mimicking properties is used to simulate the human head. The gap size variable is defined as the ratio of gap arc (L_g) length to coil arc length (L_c).

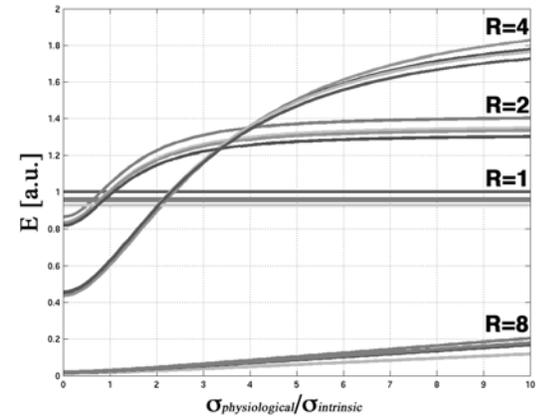


Fig. 2: The metric of performance (E) calculated using mean SNR and mean g-factor is shown for varying acceleration rates (R) and gap sizes as a function of the ratio of physiological to intrinsic noise.