Theoretical signal-to-noise penalty in parallel ultra-low-field magnetic resonance imaging

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INTRODUCTION

In ultra-low-field (ULF) magnetic resonance imaging (MRI) [1], a readily available sensory MEG array consisting of up to hundreds of SQUID detectors can be used for simultaneous acquisition of the magnetic field emitted by the precessing magnetization. Such a multichannel data acquisition immediately allows the application of the parallel MRI principle to reduce the data acquisition time, which critically depends on the number of polarization steps in ULF-MRI. In this study, we investigate the theoretical signal-to-noise ratio (SNR) penalty based on the Sensitivity Encoded (SENSE) MRI [2] at various acceleration rates using an MEG array with different subsets of pick-up coils. Our results suggest that an average geometry factor (*g*-factor) accounting for the noise amplification in 2D SENSE imaging of 1.5 or less can be achieved by an array of 102 magnetometers or 306 sensors of magnetometers and gradiometers at 16-fold acceleration. This holds the promise for 3D ULF MRI with image matrix of 64x64x64 (4-mm isotropic resolution) to be completed within nine minutes.

METHOD

We studied two types of coil array geometries. The first one is an array of 19 elements distributed over the vertex of the head. Each element of this array was designed to be a planar structure with 30-mm diameter. The geometry of each element was modeled as either a “magnetometer” (circular loop), or a “gradiometer” (planar figure-eight loop). The second array geometry was designed to allow 102 elements to cover the whole head evenly based on a 306-channel MEG system (VectorView, Elekta, Helsinki, Finland). Specifically, we used either 102 magnetometer channels (planar circular loops, denoted as array102), 102-channels of gradiometers (planar figure-eight loops, denoted as array102x or array102y), or their combination (denoted as array306) to cover the whole head. These arrays are shown in **Figure 1**

Given a coil array geometry, we used the Biot–Savart law to calculate the magnetic fields generated by a unit current on each element separately. Since our system has a measurement field oriented along the *y*-axis, the *x*–*z* plane will be the rotating plane of magnetization. Accordingly, the magnetic field components in the *x* and *z* directions consist of the real and imaginary parts of the coil sensitivity maps. Using SENSE pMRI formulation, the *g*-factor accounts for the relative SNR efficiency during accelerated scan [2]: $g_{\rho\rho}=\mathrm{sqrt}([(\mathbf{A}^H\Psi^{-1}\mathbf{A})^{-1}]_{\rho\rho}\mathrm{x}[\mathbf{A}^H\Psi^{-1}\mathbf{A}]_{\rho\rho})$. The subscript ρ indicates the voxel to be reconstructed. The encoding matrix **A** consists of the product of the aliasing operation due to sub-sampling of the *k*-space data and complex value coil sensitivity maps. Ψ is the receiver noise covariance.

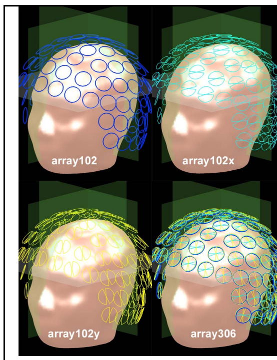

Figure 1. The geometries of using 102 or 306 pick-up coils to cover the whole head. Orthogonal slices for *g*-factor calculation are illustrated by translucent planes.

RESULTS

By increasing the array elements from 19 to 102 units, the whole head is covered more evenly, which leads to a more homogeneous distribution of the *g*-factor (**Figure 2**). Using the same array, the *g*-factor becomes worse at a higher acceleration rates. There is no clear difference between using array102, array102x, and array102y coil geometries. Using 306 sensors did not bring significant *g*-factor improvement either. Acceleration rates of 6x6 and higher make significant SNR degradation (*g*-factor > 2) at the deep brain areas. Practically, with 4x4 acceleration, most of the brain is covered with *g*-factor less than 1.5. This could be a reasonable trade-off between shortening imaging time and SNR loss.

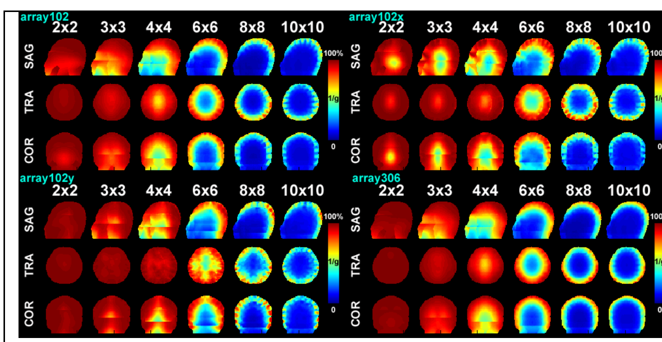

Figure 2. The spatial distribution of 1/g-facor in mid-sagittal, mid-coronal, and one axial slice using array102, array102x, array102y, and array306 geometries at R=2x2, 3x3, 4x4, 6x6, 8x8, and 10x10 respectively.

DISCUSSION

In this study, we quantitatively investigated the SNR penalty in accelerated ULF-MRI using the SENSE approach at different coil geometries. Two-dimensional accelerations factors ranging between 4 and 100 were studied. We found that 4–9-fold acceleration is possible using an array with 19 sensors. This is similar to the 3-fold acceleration achieved by a 7-channel array [3]. An array with 102 sensors covers the whole brain evenly enabling 16-fold acceleration in two dimensions with an average g-factor less than 1.5.

In practice, our calculation suggests that the parallel MRI technique can significantly shorten the imaging acquisition time for ULF-MRI . Most of the time in ULF-MRI is spent on preparing the magnetization. Suppose that one TR is 1 second, making a 3D acquisition with 64x64x64 voxels need 4,096 independent measurements. This amounts to more than 1 hour of acquisition time. Using a coil array of 102 elements, as suggested by our calculation, 16-fold acceleration can be achieved. In such a case the whole data set can be measured within 4 minutes. Zotev et al [4] have shown that with six repetitions a 3D brain volume can be imaged in 90 minutes using a 7-channel array. If our 306-channel array offered independent spatial information compared to their 7-channel array, the SNR can be improved by a factor of $\sqrt{306/7}\approx 6.6$. This SNR can be then be traded for imaging acceleration and thus a 3D volume can be acquired in 14 minutes. These numbers clearly indicate the potential significance of parallel MRI and high-density parallel detection in making ULF MRI practical.

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