Combination of Optimized Transmit Arrays and Some Receive Array Reconstruction Methods Can Yield Homogeneous Images at Very High Frequencies

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Image inhomogeneity related to high radiofrequencies is one of the major challenges for high field imaging. This inhomogeneity can be thought of as having 2 radiofrequency-field related contributors: the transmit field distribution and the reception field distribution. Adjusting magnitude and phase of currents in elements of a transmit array can significantly improve flip angle homogeneity at high field. Effective application of some wellknown parallel imaging and other receive array post-processing methods removes receptivity patterns from the intensity distribution in the final image, though noise then becomes a function of position in the final image. Here simulations are used to show that, assuming high signal-to-noise ratio, very homogeneous images in the human head can be acquired with the combination of transmit arrays and some receive array reconstruction methods at frequencies as high as 600 MHz. Magn Reson Med 54:1327-1332, 2005. © 2005 Wiley-Liss, Inc.

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Radiofrequency magnetic (\mathbf{B}_1) field inhomogeneity has been an obstacle in MRI and NMR since their inception. This problem becomes more pronounced at high radiofrequencies as wavelengths in tissue become shorter. Methods to reduce the effects of \mathbf{B}_1 field inhomogeneity have included postprocessing with (1) or without (2) prior knowledge of the B_1 field distribution, adjusting the nominal flip angle of pulses (3), use of tailored, adiabatic, and composite RF pulses (4–6), varying the coil geometry (7,8), use of dielectric padding (8,9), and adjusting the driving magnitude and phase (or impedance) of individual elements in a volume coil or coil array, previously called "RF shimming" (10–14).

Parallel imaging methods have been advanced as a method of using coil arrays during reception to reduce imaging time (15–18). An under-recognized effect of some methods of parallel MRI reconstruction is the removal of the RF coil receptivity distributions from the final image intensity distribution (19). This is true even with no acceleration of imaging (or with full *k*-space acquisition), where the resulting signal intensity distributions of these methods become similar to those resulting from phased-array

reconstruction designed to produce "uniform sensitivity" images (20).

Here simulations are used to show that very homogeneous images in the human head can be acquired at frequencies as high as 600 MHz with the combination of varying the magnitude and phase of currents within elements in a transmit array and using some receive array reconstruction techniques.

THEORY

For a single transmit/receive coil in conventional imaging, the spatial distribution of the intensity S at each location on a gradient-echo image with a long repetition time can be estimated as (21):

$$S \propto W |\sin(I|\hat{B}_1^+|\gamma\tau)|B_1^-$$
[1]

where *W* is a tissue-dependent weighting factor potentially determined by proton density, relaxation properties, and imaging sequence parameters; I is a non-dimensional value proportional to coil current magnitude; γ is the gyromagnetic ratio; τ is the **B**₁ pulse duration; and \hat{B}_1^+ and \hat{B}_1^- are the complex circularly polarized components of the \mathbf{B}_1 field rotating in the same and opposite directions of precession, respectively, and $B_1^- = |(\hat{B}_1^-)^*|$. The parameters S, W, \hat{B}_1^+ , and \hat{B}_1^- are all functions of spatial location. Here \hat{B}_1^+ and \hat{B}_1^- are calculated for unit current in the coil. Note that if \hat{B}_1^+ rotates and the nuclei precess in the counterclockwise direction defined by the usual manner using the right hand rule in a Cartesian coordinate system (22), this requires that the static magnetic (\mathbf{B}_0) field be oriented in the -z direction (23). For the B_0 field oriented in the +zdirection, \hat{B}_1^+ and \hat{B}_1^- would be interchanged in the usual equations (11,21,22).

For some parallel imaging techniques, including archetypal SENSE and archetypal SMASH, the receive coil receptivity distributions B_{1l}^- (for an *l*-element receive array) are used to encode spatial information during acquisition in such a way that in reconstruction the available signal intensity is found as a function of encoding by the applied gradient fields and the coil receptivity distributions. This available signal intensity distribution has been called "spin density" (19). Here, where we do not assume a homogeneous transmit field \hat{B}_1^+ , this could be represented as:

$$S_s \propto W |\sin(|\sum_m I_m \hat{B}^+_{1m} | \gamma \tau)|$$
 [2]

in the case where an *m*-element array is used during transmission. Here S_s is used to indicate intensity on an image

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using SENSE or SMASH reconstruction. Mathematically, the removal of the receptivity distribution from the final image for archetypal SENSE and SMASH is made clear by Sodickson and McKenzie (19), and graphically it is clear in the original publication of the SENSE technique (16), where for a reduction factor of 1 (no acceleration), an image with a commonly used (sum of squares) phased array reconstruction method producing images heavily weighted by receptivity is juxtaposed with an image with SENSE reconstruction. Other phased array reconstruction techniques designed to create a "uniform sensitivity image" (20) likewise effectively remove the receptivity distribution from the final image so that S_s could similarly be used to indicate image intensity.

In this treatment we have taken a very simple approach in order that the basic concepts be transparent, but this simple approach also requires a fair number of assumptions and concomitant caveats. The accuracy to which the "spin density" at each location can be calculated in practice depends on the accuracy of the knowledge of the coil receptivity distributions B_{1l}^{-} in SENSE, and depends on the accuracy with which combinations of these receptivity distributions can be combined to fit pre-determined spatial harmonics in archetypal SMASH. Also, the noise distribution in parallel imaging becomes a function of location, and this becomes increasingly problematic at higher accelerations (16). There are also many nontrivial hardware issues, including coil decoupling, that must be dealt with for these methods to work effectively. With respect to all these important factors, here we intend only to make a relatively simple but potentially very important observation: ideally, some coil array reconstruction techniques, including some commonly used parallel imaging methods, have the effect of removing the weighting by the receptivity distribution from the final image.

This is not, however, the case for all parallel imaging methods. In PILS (17), for example, magnitude images from different coils are summed, and aliasing from different coils is avoided as the acceleration is increased by maintaining an acquisition field of view that is larger than the receptivity distribution of the individual coils. Also, in GRAPPA (18), where acquired lines of k-space from different coils are combined in an empirically determined manner to fill remaining lines of *k*-space for all coils, there is currently no implicit mechanism for removal of the coil receptivity distributions from the final image. We cannot here discuss all currently existing methods of array reconstruction, nor anticipate methods that may be used in the future, but the distinction that some methods remove this receptivity distribution from the final image while others do not is important in the remainder of this work, and has not, to our knowledge, been made previously with regard to parallel imaging methods.

METHODS

An anatomically accurate, multi-tissue human head was modeled within a 16-element, elliptical, stripline coil array. Each element of the array was modeled as a 2 cm-wide and 15 cm-long thin strip of copper oriented in the longitudinal direction and placed with equal spacing on the surface of an ellipse with a short (left-right) axis of 21 cm and a long (anterior-posterior) axis of 24 cm. Both ends of each element were connected (with thin wires in a roughly radial direction) to an elliptical shield of copper with a 15 cm length and short and long axes of 23 cm and 27 cm modeled concentrically with the array. The elliptical shape was chosen to conform better to the shape of the head than would a circular shape, as coil arrays are often designed to conform to the anatomy of interest. Longitudinal slots of 1 cm width were placed in the shield between each pair of neighboring elements. One current source was placed in each connecting wire with sources at opposite ends of a given element having equal magnitudes and phases and being oriented in opposite directions. The head model was created with methods described previously (21) at a resolution of 5 mm. The head model within the coil array with array elements labeled for reference is shown in Fig. 1. The finite-difference time-domain method was used to model the field produced by each element driven individually using the finite difference time domain method at 300, 400, 500, and 600 MHz. Electrical properties of tissues at these frequencies were obtained from the literature (24). All FDTD calculations were performed with the aid of commercially available software ("xFDTD"; Remcom, Inc.; State College, PA). Values of \hat{B}_1^+ and \hat{B}_1^- were calculated from the results of the FDTD calculation using methods described previously (21).

The results of the field calculations for all individual coils at each frequency were utilized in home-built simple optimization routines using Matlab (The MathWorks; Natick, MA). Starting with current magnitudes and phases roughly like those that would result from quadrature excitation in a volume coil, the magnitudes and phases of the individual coil currents were varied incrementally and sequentially with the goal of improving the homogeneity of S_s , calculated as in Eq. [2], with W = 1 in brain (white matter, gray matter, and CSF, and not including cerebellum or brain stem) and W = 0 elsewhere. In Eq. [2], variation of magnitude of the *m*-th coil current could be achieved by varying I_m , while varying the phase of the *m*-th coil current can be accomplished through variation of the phase of \hat{B}_{1m}^+ . By using Eq. [2] here, we make use of the fact that with perfect knowledge of the field patterns and sample geometry (as we have in calculations), the signal intensity distribution resulting from reconstruction algorithms like those used in SENSE can be calculated quite simply, though in experiments this will produce a nonuniform noise distribution not considered here. This optimization was performed both on a single axial plane passing roughly through the center of the ventricles at 300, 400, 500, and 600 MHz, and for the whole brain volume at 600 MHz. In optimizing the homogeneity of S_s on the single plane, the ratio of the sum of the squares of the differences between the local values of S_s and the mean value on the plane to the mean value was minimized. This ratio, rather than just the sum of the squares, was minimized to avoid the perfectly homogeneous case of no fields anywhere (no current in any elements) and to ensure overall signal magnitude is maximized in the solution. For the whole brain optimization, if a difference between a local value of S_s and the mean value in the brain was less than 20% of the mean value, its square was not included in the sum, allowing greater flexibility in values for S_s within 20% of the mean value.





Given an initial assignment of 16 current magnitudes and phases (32 variables) and initial step size of 10% change in magnitude and 60° in phase change, each optimization was performed by examining the homogeneity for a change in each of the 32 variables sequentially, immediately changing to the new value if it showed improvement. This was repeated until changes in none of the 32 variables improved the homogeneity, at which time the step size would be changed to negative one-half of its current value and the process would be repeated. When the step size reached a value smaller than 0.1% change in magnitude and 1° change in phase, the optimization routine was ended. Although routines such as this can settle at local optima rather than global, in our experience with these problems, starting from several very different sets of magnitude and phase generally resulted in the same optimal result, indicating it likely to be the global optimum.

RESULTS

Figure 2 shows the current magnitude and phase in each element before (original) and after optimization on the single axial plane at 300, 400, 500, and 600 MHz, and after optimization in the whole brain at 600 MHz (600 w.b.). All phase values are shifted so that the phase of element 1 is zero in all cases. It is interesting that the optimal currents at 600 MHz more closely resemble those of the original configuration than do the optimal currents at lower frequencies. Otherwise, the numerically optimized current distributions for this sample and coil geometry do not

appear to exhibit any obvious symmetry or easily understandable trends.

Figure 3 shows the distribution of image intensity S_s at 300, 400, 500, and 600 MHz before and after optimization in the brain on the single axial plane shown. After finding optimal coil current magnitudes and phases, the image intensity distribution is very homogeneous at each frequency. Even at 600 MHz, all values for S_s on the plane shown are within 10.6% of the mean after optimization.

Figure 4 shows the distribution of image intensity S_s on mid-axial, mid-coronal, and mid-sagittal planes at 600 MHz before and after optimization in the whole brain. Again, the intensity distribution is much more homogeneous after finding optimal coil magnitudes and phases. With the original configuration, intensities in large regions of the brain are from below 50% (black) to above 120% (orange) of the mean value. While the intensity on the mid-sagittal plane in the original configuration may appear homogeneous by itself, much of this is over 120% of the mean value in the brain. After finding optimal coil magnitudes and phases, the vast majority of the brain volume has $S_{\rm S}$ within 20% of the mean (royal blue to yellow), with only a small region near the center deviating by more than 50% from the mean. In this region the strength of $|\sum_m I_m \hat{B}_{1m}^+|$ is greater than that elsewhere, resulting in a flip angle approaching 180°.

DISCUSSION

Initial indications are that the combination of a transmit coil array and application of some parallel reconstruction



FIG. 2. Magnitudes (left) and phases (right) of element currents before (original) and after optimization on the single axial plane at 300, 400, 500, and 600 MHz, and after optimization in the whole brain at 600 MHz (600 w.b.). All phase values are shifted so that the phase of element 1 is zero in all cases.



FIG. 3. Image intensity using low-acceleration archetypal SENSE or SMASH reconstruction (S_s) distributions for head in 16element array at 300, 400, 500, and 600 MHz before and after optimization of image homogeneity on plane shown by variation of magnitude and phase of currents in transmit coils. Scale gives fraction of mean intensity value on plane shown. Values less than 50% of the mean value appear as 50% (black). Even at 600 MHz, all S_s values on the plane are within 10.6% of the mean after optimization.

techniques can yield very homogeneous images in the head at very high frequencies. This takes advantage of a lesser-recognized particular characteristic of some parallel imaging techniques, namely, the removal of the coil receptivity distribution from the final image.

It appears that lower current magnitudes are required to produce optimal field patterns (taking flip angle into account) at higher frequencies (Fig. 2). This does not by itself mean that lower power would be required at the higher frequency, as both the impedance of the elements and the power absorption by the sample for a given total RF magnetic field pattern are generally expected to increase with frequency. Previous calculations for power absorption as a function of frequency for other field distributions in various representations of a human head have generally not gone as high as 600 MHz (21,25,26), but some (considering linear excitations and not considering circularly polarized field components) have indicated a possible decrease in required power somewhere above 300 MHz (25,26). The current phases and magnitudes in the optimal configurations here, however, are very unlike those from previous analyses of power, especially at lower frequencies. A full analysis of power requirements will require significant further analysis, including an evaluation of the electrical fields throughout the sample in the optimal drive configuration.

It is interesting to note that in the original configuration (before optimization), the effect of central brightening does not become more accentuated as we approach 600 MHz from 300 MHz, but rather becomes less distinct (Fig. 3). This is due to the non-axis-symmetric geometry of both the coil and the sample in these simulations. Certainly for a spherical or cylindrical sample concentric within a cylindrical coil driven with axis-symmetric magnitudes and phase evolving with azimuthal angle, we would expect nearly perfect constructive interference at the center while wavelengths decreased with increasing frequency, resulting in increasing degree of destructive interference and continued distinctness of the relative brightness at the center. Related to this, the optimal configurations at

Sagittal Coronal Axial

FIG. 4. Distribution of S_s on mid-sagittal, midcoronal, and mid-axial planes at 600 MHz before and after optimization in the whole brain. The color scale is the same as in Fig. 3. After finding optimal coil magnitudes and phases, the vast majority of the brain volume has S_s within 20% of the mean, with only a small region near the center deviating by more than 50% from the mean. slightly lower frequencies where wavelength effects are still important tend to have magnitude and phase distributions (Fig. 2) that will result in slightly asymmetric optimized field patterns (Fig. 3) to avoid strong constructive interference at the center. At all frequencies the optimized phases appear to increase in a nearly linear fashion going from element to element, but at a slope approximately twice as large at the lower frequencies. This results in currents around the coil passing through 2 whole phase cycles rather than just 1, resulting in 4 locations of strong current at any one time as opposed to just 2 in a typical quadrature volume coil. While this produces a B_1 field magnitude distribution that is relatively weak near the center of the array, because the field at the center tends to have more perfect circular polarization, B_1^+ ends up being quite homogeneous.

The results in this work for optimizations on a single axial plane (Fig. 3) appear more homogeneous than results of some previous attempts to find optimal current configurations at high field (12,13), partly because here we are plotting and optimizing the homogeneity of the signal intensity, rather than just a B_1 field magnitude. This results in optimal flip angles near 90°, where a relatively large variation in B_1^+ can result in a relatively small variation in the sine of the flip angle. Results of optimizations requiring other nominal flip angles (such as lower flip angles for gradient-echo sequences with short TR) with the techniques used here would likely not yield as homogeneous signal intensity distributions.

Importantly, previous attempts to optimize image homogeneity using techniques similar to those shown here but with mathematical representations that did not remove the receptivity distribution (27) were not nearly as successful as the results shown here using Eq. [2]. This indicates that optimizing image homogeneity using transmit arrays and reconstruction methods that remove the receptivity distribution from the final image will be much more successful than using transmit arrays and other reconstruction techniques.

It has been shown that parallel imaging techniques themselves will likely become more effective in high field imaging (28). The success of many parallel imaging techniques, including archetypal SENSE, depends on the ability to accurately measure the receptivity distribution of each coil used in reception. This is most easily accomplished with a homogeneous flip angle throughout the imaging region, so coupling transmit arrays (to achieve homogeneous flip angle distributions) with parallel imaging methods may also further improve the ability to effect parallel imaging methods at high field strengths.

The solution arrived at for optimal signal homogeneity for the entire brain (Fig. 4) is notably darker at the center of the brain, though the original configuration tends to have a bright region in the center. In fact, this is due to flip angles approaching 180° at the center of the brain here: the transmit field strength is still higher than in the surrounding regions though the image intensity is weaker. The achievable homogeneity throughout the whole brain is not as great as that on a single plane because of the nature of Maxwell's equations in 3-dimensional space (11). Still, with more elements than simulated here, including a greater number of shorter elements distributed in the longitudinal direction, we anticipate that greater homogeneity than realized here in the whole-brain volume should be possible. It is also conceivable that in multi-slice (rather than true 3-dimensional) acquisitions, magnitudes and phases optimized for the excitation of each individual slice could be applied separately at the appropriate times, allowing for optimized image intensity distributions more like those in Fig. 3 to be produced throughout the brain. Although this could confound the application of parallel imaging reconstruction methods as currently formulated to the whole volume, they could still be applied on the slices individually.

Overall, the method for improving image homogeneity here could be described as a combination of using postprocessing with some foreknowledge of the field distribution (e.g., "SENSE" or "SMASH") coupled with use of a transmit array with optimized current magnitudes and phases (sometimes called "RF shimming" or "B1 shimming" previously). This combination of methods may not be capable of producing as homogeneous flip angle distributions as adiabatic or tailored RF pulses, but should require shorter and simpler RF pulses and less RF power be adsorbed into the tissues, which is also an issue of concern in high field MRI. The methods proposed here could potentially be electronically controlled in the future, thus being more versatile, flexible, and more rapidly optimized for differing subjects than methods using dielectric padding or based on changing the transmit coil geometry.

Technological advances to implement a transmit array with independently controlled current or voltage sources are currently being developed for MRI. Previously, RF shimming has been accomplished by varying the impedance of individual elements in a quadrature-driven TEM coil (10). Work is underway toward accomplishing RF shimming under algorithm-driven electronic control using several parallel RF voltage amplifiers at frequencies as high as 400 MHz in the human head (29). Recently, implementation of a transmit system with independently controlled element currents (more like that simulated here) was facilitated with voltage-controlled current sources to drive the desired currents directly in individual elements (30).

In conclusion, while RF-related homogeneity issues have been a concern throughout the history of NMR and MRI, here we demonstrate that by further developing and combining some existing techniques, this problem may have at least one reasonably good solution in head imaging at frequencies as high as 600 MHz.

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