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

Christopher M. Collins,* Wanzhan Liu, Bryan J. Swift, and Michael B. Smith

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 well-known 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 (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 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):

\[ S = W|\sin(\gamma B_1^* B_{1z})| \]  

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 B_1^* and B_{1z} are the complex circularly polarized components of the B_1 field rotating in the same and opposite directions of precession, respectively, and B_{1z} = |[B_1^*]|. The parameters S, W, B_1^*, and B_{1z} are all functions of spatial location. Here B_{1z} and B_{1*} are calculated for unit current in the coil. Note that if B_{1z} 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 (B_0) field be oriented in the –z direction (23). For the B_0 field oriented in the +z direction, B_{1z} and B_{1*} would be interchanged in the usual equations (11,21,22).

For some parallel imaging techniques, including archetypical SENSE and archetypical SMASH, the receive coil receptivity distributions B_{1r} (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 B_{1r}, this could be represented as:

\[ S_a = W|\sin(\pi \sum_{m=0} B_{1r,m}^* |\gamma\tau|) \]  

in the case where an m-element array is used during transmission. Here S_a is used to indicate intensity on an image.
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 $S_e$ 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 $R_{ij}$ 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 $B_m^e$ and $B_m^s$ 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 $S_m^e$, 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 $B_m$. 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 $S_m^e$ on the single plane, the ratio of the sum of the squares of the differences between the local values of $S_m^e$ 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 $S_m^e$ 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 $S_m^e$ 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 $S_i$ 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 $S_i$ on the plane shown are within 10.6% of the mean after optimization.

Figure 4 shows the distribution of image intensity $S_i$ 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 $S_i$ 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 $\sum L_i B_{\text{max}}^i$ 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
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

FIG. 3. Image intensity using low-acceleration archetypal SENSE or SMASH reconstruction ($S_s$) distributions for head in 16-element 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.

FIG. 4. Distribution of $S_s$ on mid-sagittal, mid-coronal, 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.
sults in optimal flip angles near 90°, where a relatively
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of some previous attempts to find optimal current config-
axial plane (Fig. 3) appear more homogeneous than results
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 config-
urations at high field (12,13), partly because here we are
plotting and optimizing the homogeneity of the signal
ensity, rather than just a B1 field magnitude. This re-
results in optimal flip angles near 90°, where a relatively
large variation in B1 can result in a relatively small varia-
tion in the sine of the flip angle. Results of optimizations
quiring other nominal flip angles (such as lower flip an-
gles for gradient-echo sequences with short TR) with the
techniques used here would likely not yield as homoge-
neous signal intensity distributions.

Importantly, previous attempts to optimize image homo-
genity 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 distribu-
tion from the final image will be much more successful
than using transmit arrays and other reconstruction tech-
niques.

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 tech-
niques, including archetypal SENSE, depends on the abil-
ity to accurately measure the receptivity distribution of
each coil used in reception. This is most easily accom-
plished with a homogeneous flip angle throughout the
imaging region, so coupling transmit arrays (to achieve
homogeneous flip angle distributions) with parallel imag-
ing 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 trans-
mittance field strength is still higher than in the surrounding
regions though the image intensity is weaker. The achieve-
izable 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 lon-
gitudinal 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 post-
processing with some foreknowledge of the field distribu-
technique (e.g., “SENSE” or “SMASH”) coupled with use of a
transmit array with optimized current magnitudes and
phases (sometimes called “RF shimming” or “B1 shim-
ming” previously). This combination of methods may not
be capable of producing as homogeneous flip angle distribu-
tions as adiabatic or tailored RF pulses, but should
require shorter and simpler RF pulses and less RF power
to 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 opti-
mized 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 imped-
ance 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, imple-
mentation of a transmit system with independently con-
trolled 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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