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Simple Analytical Dual-Band Spectral-Spatial RF Pulses for B1+ and Susceptibility Artifact Reduction in Gradient Echo MRI

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Abstract

Susceptibility artifacts and transmission radio frequency (RF) field (B1+) inhomogeneity are major limitations in high field gradient echo MRI. Previously proposed numerical two-dimensional (2D) spectral-spatial RF pulses have been shown to be promising for reducing the through-plane signal loss susceptibility artifact by incorporating a frequency dependent through-plane phase correction. This method has recently been extended to four-dimensional (4D) spectral-spatial RF pulse designs for reducing B1+ inhomogeneity as well as the signal loss. In this manuscript, we present simple analytical pulse designs for constructing 2D and 4D spectral-spatial RF pulses as an alternative to the numerical approaches. The 2D pulse capable of exciting slices with reduced signal loss and is lipid suppressing. The 4D pulse simultaneously corrects signal loss as well as the B1+ inhomogeneity from a body coil transmitter. The pulses are demonstrated with simulations and with gradient echo phantom and brain images at 3T using a standard RF body coil. The pulses were observed to work well for multiple slices and several volunteers.

Keywords

Multi-dimensional RF pulses; spectral-spatial RF pulses; dual-band RF pulse; susceptibility artifacts; and B1+ inhomogeneity

INTRODUCTION

High field (B0) MRI systems have advantages including increased signal-to-noise ratio (SNR) (1) and increased contrast for Blood-Oxygen-Level-Dependent functional MRI (fMRI) (2). Unfortunately, several confounds need to be addressed before these advantages can be fully realized. One confound is susceptibility artifacts that appear as distortions or as large voids in the images. They are particularly problematic in gradient echo imaging applications with a long echo time (TE) such as fMRI. The through-plane signal-loss artifact in axial slices is of primary concern due to the close proximity of air/tissue boundaries to the inferior brain areas. A second problem is artifacts produced by inhomogeneity in the transmitted radiofrequency (RF) field (B1+) (3). These appear as regions of increased and decreased image magnitude at 3T.

Numerous methods have been proposed to mitigate the susceptibility induced signal loss artifact. These include z-shim methods (4), thin slice averaging (5), shim coils (6), multi-echo sequences (7), parallel transmission (8), and tailored RF pulses (9). Methods that can

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reduce signal loss in a single excitation are of particular importance because they do not increase scan time. Numerically designed two-dimensional (2D) spectral-spatial pulses have recently been shown to be capable of reducing the signal loss artifact in a single excitation (10). The idea behind the use of spectral-spatial excitations is that regions needing the through-plane pre-compensatory phase will also be off resonance. The spectral-spatial pulse can then be designed to only phase compensate these off-resonant spins and leave the on-resonance spins unaffected.

Numerous methods have also been proposed to mitigate B1+ inhomogeneity artifacts. These include specially designed coils (11), adiabatic pulses (12), “B1+ shimming” using multiple transmitters (13), and small-tip-angle multi-dimensional tailored RF pulses (14). Multi-dimensional RF pulses are advantageous because they allow for the in-plane slice uniformity to be spatially tailored. Previous work shows that a three-dimensional (3D) “fast- k_z ” tailored RF pulse can reduce the B1+ inhomogeneity from a body coil transmitter (15). The fast- k_z design is very promising because it is capable of exciting thin slices with short pulse lengths in a single excitation. The fast- k_z pulse has also been demonstrated for multiple transmitter use (13).

The fast- k_z and 2D spectral-spatial pulses can be combined to form a four-dimensional (4D: x, y, z , and f) spectral-spatial pulse to reduce both B1+ inhomogeneity and signal loss. This pulse may be useful in applications such as fMRI at ultra-high field where both types of artifacts are a concern. The 4D spectral-spatial pulse concept was recently demonstrated using a numerical design for multiple transmitters (16). A drawback of the 4D (as well as the 2D) spectral-spatial pulse design is the need to numerically solve a system of linear equations. This can take several minutes on a standard (i.e. dual 2.8 GHz quad-core) computer, which can become cumbersome if several pulses need to be designed.

We propose a dual-band k -space approach for designing both 2D and 4D spectral-spatial pulses. This approach is analytical and simpler than the previously proposed numerical designs, requiring under a second to construct on a dual 2.8 GHz quad-core computer. The 2D pulse reduces through-plane signal loss due to susceptibility variations and is lipid suppressing as well. The 4D pulse simultaneously mitigates the B1+ inhomogeneity from a body coil transmitter and the through-plane signal loss. The pulses are validated with simulations as well as with phantom and human imaging experiments at 3T. The pulses are shown to be effective at reducing these artifacts in human brains using a gradient echo sequence and a transmit body coil.

THEORY

DUAL-BAND SPECTRAL-SPATIAL PULSES

The small tip-angle approximation provides an intuitive method for RF pulse design using the concept of excitation k -space (17). In this picture, one can think of the excited magnetization $M(z, f)$ as being the Fourier transform of an RF pulse $B(t)$ of length T :

$$M(z, f) = \int_0^T B(t) e^{-ik_z(t)z - ik_f(t)f} dt, \quad [1]$$

$$\text{where } k_z(t) = \gamma \int_t^{T_s} G_z(s) ds \quad \text{and} \quad k_f(t) = T - t. \quad [2]$$

A 2D spectral-spatial pulse can be designed using a train of standard slice-select sub-pulses each of shape $p(t)$ and length T_z (18). The slice-select gradient G_z is typically a series of trapezoids of length T_z with alternating polarity. The excited magnetization for a 2D spectral-spatial pulse can be written as

$$M(z, f) = P(z) W(f) * \sum_{n=-\infty}^{\infty} \delta(f - n f_s). \quad [3]$$

In this equation f_s is the separation between spectral aliasing bands and $P(z)$ is the Fourier transform of $p(t)$ and has a width (slice thickness) of z_0 . The function $W(f)$ is the Fourier transform of a temporal weighting function that is applied along the entire pulse length. The width of $W(f)$ defines the frequency pass-band, f_p . A fly-back design is often used where $p(t)$ is applied only on the positive G_z lobes. The advantage of the fly-back design is that it is more robust to timing errors that can result from eddy currents, for example (19). The separation between spectral aliasing bands is $f_s = 1/2T_z$ for a fly-back pulse.

If it is desirable to excite unique slices centered at two different frequencies, a dual-band spectral-spatial pulse can be constructed as the sum of two fly-back pulses of length T (20):

$$B(t) = B_+(t) + B_-(t) e^{-i\Delta f(t-T)}. \quad [4]$$

The pulses $B_{\pm}(t)$ represent fly-back spectral-spatial pulses located on the positive and negative G_z lobes. The frequency modulation of $B_-(t)$ shifts its pass-band by Δf while the pass-band of $B_+(t)$ remains centered on resonance. The excited magnetization profile can be written as

$$M(z, f) = P(z) W(f) * \sum_{n=-\infty}^{\infty} [\delta(f - n f_s) + \delta(f - \Delta f - n f_s)]. \quad [5]$$

SPECTRAL-SPATIAL PULSES FOR REDUCED SUSCEPTIBILITY ARTIFACT

Previous studies have demonstrated that a 2D spectral-spatial pulse can reduce the signal loss susceptibility artifact. This requires the assumption that the susceptibility induced through-plane phase is frequency dependent:

$$\phi(z, f) = \gamma G_s(f) z T E. \quad [6]$$

Here $G_s(f)$ is the susceptibility gradient that is assumed to be a function of frequency. The pulses can be designed using a numerical approach starting with the desired magnetization

$$M(z, f) = P(z) e^{-i\phi(z, f)} \quad [7]$$

and solving for $B(t)$ using Eq. [1]. The susceptibility gradient is modeled as a linear function of frequency: $G_s(f) = \alpha f$, where α is on the order of $-1.0 \mu\text{T/m/Hz}$.

We propose that a similar pulse can be designed analytically using the dual-band approach discussed above. Phase pre-compensation at Δf can be introduced by time shifting $B_-(t)$ an amount $\Delta t = G_s T E / G_z$:

$$B(t) = B_+(t) + B_-(t - \Delta t) e^{-i\Delta f(t-T) + i\phi_0}. \quad [8]$$

Fourier transformation of $B(t)$ gives the excited magnetization profile:

$$M(z, f) = p(z) w(f) * \sum_{n=-\infty}^{\infty} [\delta(f - n f_s) + \delta(f - \Delta f - n f_s) e^{-i\gamma G_z z \Delta t + i\phi_0}]. \quad [9]$$

The term φ_0 is an adjustable constant phase that is added to match the phases between the two bands where they overlap.

Figure 1 (a) shows an example of a dual-band 2D spectral-spatial pulse. The rows from top to bottom are the real and imaginary parts of the RF and the z -gradient waveforms. The black line shows $B_+(t)$ and the gray line shows $B_-(t)$ in the top two rows. Note that $B_-(t)$ has a phase modulation as well as a time shift compared to $B_+(t)$. Figure 2 (a) shows a Bloch simulation of the magnitude and phase of $M(z, f)$ excited by this pulse, evaluated along the slice-select and frequency directions. Note the two bands centered at 0 and 125 Hz. The band at 0 Hz has no through-plane pre-compensatory phase whereas the band at 125 Hz has a linear phase of approximately 2π radians across the slice. There is also no magnetization at -440 Hz where lipid resonates, making the pulse fat suppressing as well. The pulse parameters are discussed in more detail in the Methods section below.

FAST- k_z PULSES FOR REDUCED B1+ INHOMOGENEITY ARTIFACT

The 3D fast- k_z pulse can be used to compensate the in-plane B1+ inhomogeneity produced by a body coil at 3T (15). The pulse is constructed as a series of slice-select sub-pulses similar to a 2D spectral-spatial pulse except they are separated by x - y gradient “blips” to weight discrete points in k_x - k_y . A “five-rung” pulse can be designed using five k_x - k_y points weighted by A_0 and A_1 .

$$h(k_x, k_y) = A_0 \delta(k_x, k_y) + A_1 \left[\delta(k_x + k_0, k_y) + \delta(k_x - k_0, k_y) + \delta(k_x, k_y + k_0) + \delta(k_x, k_y - k_0) \right]. \quad [10]$$

If k_0 is set to be small compared to the inverse of the field of excitation (FOX), Fourier transformation of $h(k_x, k_y)$ gives an in-plane magnetization with a quadratic profile:

$$H(x, y) \cong A_0 + 4A_1 - A_1(2\pi k_0)^2 (x^2 + y^2). \quad [11]$$

This profile will compensate the B1+ inhomogeneity of a body coil, which is typically observed as increased magnitude or brightness in the center of the image. If $k_0 = 1 / 2\text{FOX}$ and $A_0 = 1$ then $A_1 = \varepsilon / (4\varepsilon + (\pi / 2)^2)$ where ε is the fractional difference between the center and the edge of the excitation profile. Values of ε equal to 0.5 to 1.0 have previously been found to work well for human brain imaging using a body coil at 3T.

4D PULSES FOR REDUCED SUSCEPTIBILITY AND B1+ ARTIFACT

A dual-band 4D spectral-spatial pulse that simultaneously compensates for through-plane signal loss and B1+ inhomogeneity can be made by replacing the pulses in $B_+(t)$ and their corresponding gradients with fast- k_z pulses each of length T_{3D} . The frequency separation between aliasing lobes is then $f_s = 1 / (T_{3D} + T_z)$. If the series of fast- k_z pulses is represented by $B_{3D}(t)$, then the pulse can be written as

$$B(t) = B_{3D}(t) + B_-(t - \Delta t) e^{-i\Delta f(t - T)}. \quad [12]$$

Fourier transformation of $B(t)$ gives the excited magnetization profile:

$$M(z, f) = P(z) W(f) * \sum_{n=-\infty}^{\infty} \left[H(x, y) \delta(f - nf_s) + e^{i\gamma G_z z \Delta + i\phi_0} \delta(f - \Delta f - nf_s) \right]. \quad [13]$$

This pulse will compensate the magnetization that is on resonance for B1+ inhomogeneity and refocus regions off resonance by Δf for through-plane signal loss. We are assuming that B1+ inhomogeneity compensation is not needed for the off resonant regions because the

susceptibility artifacts are localized to small regions. This is a reasonable assumption given the slow spatial variation and small magnitude of the inhomogeneity at 3T.

Figure 1 (b) shows an example of a dual-band 4D spectral-spatial pulse. The rows from top to bottom are the real and imaginary parts of the RF and the x -, y -, and z -gradient waveforms. In the top two rows, the black line shows $B_{3D}(t)$ and the gray line shows $B_{-}(t)$. The dashed box shows the location of a single five-rung fast- k_z 3D pulse for reference. Like the 2D spectral-spatial pulse, $B_{-}(t)$ has a phase modulation as well as a time shift to correct the signal loss artifact. Figure 2 (b) shows a simulation of the magnitude and phase of $M(z, f)$ excited by the pulse evaluated along the slice-select and frequency directions. Similar to the 2D pulse, there are two pass bands centered at 0 and 125 Hz. The band at 0 Hz has the desired quadratic x - y slice profile to correct for B1+ inhomogeneity and the band at 125 Hz has a linear phase of approximately 2π radians across the slice to reduce signal loss. The parameters are discussed in more detail below.

METHODS

The spectral-spatial pulses were designed using custom Matlab (The Mathworks, Natick, MA) programs running on a dual 2.8 GHz Quad-Core Macintosh Pro (Apple, Cupertino, CA) workstation. The peak gradient was 30 mT/m and the peak slew rate was 180 T/m/s. The 2D spectral-spatial pulse was designed using 18 alternating Gaussian slice-select sub-pulses $p(t)$ with a $z_0 = 5$ mm slice-thickness and an excitation resolution of 1.8 mm. The length of each pair of sub-pulses was $2T_z = 1.3$ ms giving $f_s = 770$ Hz, which is large enough to suppress the lipid signal at -440 Hz at 3T. The time envelope was Gaussian giving a pass-band f_p of approximately 100 Hz. Signal loss phase pre-compensation parameters of $\Delta f = 125$ Hz and $\Delta t = -10$ μ s were determined by *post-hoc* visual inspection of corrected brain images in preliminary experiments, which corresponded to $\alpha = -0.8$ μ T/m/Hz. The total length of the pulse was approximately 12 ms including a small z -gradient rewinder. Figure 1 (a) shows a diagram of the 2D spectral-spatial pulse as discussed above.

The 4D spectral-spatial pulse was designed using five-rung fast- k_z 3D pulses as described above. Each slice-select sub-pulse in the fast- k_z pulse had a 3.6 mm excitation resolution and 5 mm slice thickness. The length of each fast- k_z pulse was $T_{3D} = 2.2$ ms and the in-plane FOV was 22 cm. A 0.65 ms long slice-select pulse followed each fast- k_z pulse and was identical to that used in the 2D design. Each pair was repeated four times giving a total pulse length of approximately 12 ms including a rewinder. The time envelope was Gaussian with a pass band f_p of approximately 100 Hz. The separation between aliasing lobes was approximately 350 Hz, which is too small for lipid suppression. Signal loss phase pre-compensation parameters of $\Delta f = 125$ Hz and $\Delta t = -10$ μ s were used as well. The degree of B1+ correction $\varepsilon = 0.5$ was determined by visual inspection in preliminary experiments. No further optimization of the in-plane slice profile was performed. Figure 1 (b) shows a diagram of the 4D spectral-spatial pulse, as discussed above.

Phantom scans were first conducted to validate the pulse designs. The imaging was performed on a Siemens (Siemens Medical Systems, Erlangen, Germany) TIM Trio 3T whole body scanner with a body coil. A standard 2D (Fast Low Angle SHot) FLASH sequence for gradient echo imaging was used (30/1000 ms TE/TR, 60° flip angle, eight 5 mm thick slices, 1 mm slice gap, 22 cm FOV, 128 $\times$ 128 matrix size, 2:08 scan time). A lipid and water phantom, constructed from one bottle of vegetable oil bundled with three bottles of water, was used to demonstrate the effectiveness of the lipid suppression of the 2D spectral-spatial pulse. The improvement in B1+ homogeneity was measured using a standard spherical water phantom. Maps of B1+ and B1- for the body coil were obtained using a

multi-angle method (21). Normalized B1⁺ maps were divided from the phantom images to account for receiver inhomogeneity.

Human brain images were then obtained on five healthy adults scanned after informed consent approved by the University of Hawaii and Queens Medical Center Internal Review Board. The scanning protocol was to first apply a three-plane localizer to align the eight gradient echo axial slices around the sinus region. A set of gradient echo images was then acquired with minimum TE for comparison purposes. Then a set of images with TE = 30 ms was acquired to show the signal loss susceptibility artifact. These two sets of images were excited with a standard slice-select pulse with the same profile, $p(t)$, as the spectral-spatial pulses. Two sets of images were then acquired at a 30 ms TE using the 2D and 4D spectral-spatial pulses. Finally, images were acquired at a 30 ms TE using the previously reported numerical 2D spectral-spatial pulse design with $\alpha = -0.8 \mu\text{T/m/Hz}$ (10). The effects of the receiver inhomogeneity B1⁺ were ignored in the human scans.

RESULTS

Figure 3 shows images of the oil and water phantom acquired at a 10 ms TE. This was the shortest echo time possible for the spectral-spatial pulses. Figure 3 (a) is the control image acquired with the standard pulse showing images of both water and lipid. The image in (b) was acquired with the standard pulse using fat saturation and (c) shows an image acquired with 2D spectral-spatial pulse. Both of these images show reduced lipid magnitude compared to the control image. Regions of interest (ROIs) were selected in one water bottle and the oil bottle as approximately indicated by the dashed rectangles. The sum of the magnitude in these ROIs was normalized to that of water value and displayed by the bar graph under each image. The lipid/water magnitude ratios are 0.42, 0.052 and 0.047 for (a), (b), and (c).

Figure 4 (a) shows images of the spherical water phantom acquired at a 10 ms TE. The upper left image was acquired with the standard pulse. The B1⁺ inhomogeneity from the body coil can be seen. The upper right, lower left, and lower right are images acquired with the 4D spectral-spatial pulse using increasing degrees of B1⁺ inhomogeneity compensation $\epsilon = 0.5, 1.0, \text{ and } 1.5$. The normalized standard deviation of the pixel magnitudes in each image after masking out background noise was 0.109, 0.069, 0.036, and 0.025 respectively. Plots of the magnitude profile from left to right across each phantom image are shown in Fig. 4 (b).

Figure 5 shows two representative brain slices from one of the five subjects. The top row (a) shows a set of control images acquired at a 10 ms TE using the standard pulse. The second row (b) shows a set of identically acquired images at a 30 ms TE. Note the signal loss artifacts due to susceptibility variations in (b). Rows (c) and (d) show the same slices acquired with a 30 ms TE using the 2D and 4D spectral-spatial pulses. Visual inspection of these slices indicates that both pulses have recovered signal loss. The images acquired with the 4D spectral-spatial pulse are also more uniform and absent of the bright center typically seen at 3T. Row (e) shows the same slices acquired with the numerical 2D spectral-spatial pulse. Visual inspection shows no difference between the signal recovery in these images and that of the analytical 2D pulse shown in (c). Similar results were seen in the other four subjects and other slices.

Quantitative analysis of the signal loss recovery was performed in ROIs taken in the orbitofrontal area in all slices in all subjects where signal loss was observed. The recovery percentage of pixels with magnitude below 20% of the normalized ROI maximum from the standard pulse was chosen as an indication of susceptibility artifact correction. In this

manner, pixels outside of the artifact-affected region in the ROI were excluded. The pulses recovered 100% of the lost signal in the upper slices, but were less effective in lower slices, resulting in an average percent signal recovery of 68.1 and 71.5 with p-values of 2.1×10^{-5} and 2.9×10^{-4} for the 2D and 4D pulses, respectively. The normalized standard deviation across the whole brain was selected as an indication of B1+ homogeneity. The average percent decrease in the normalized standard deviation was 12.6 with a p-value of 6.5×10^{-4} for the 4D pulse.

DISCUSSION AND CONCLUSIONS

In conclusion, we presented a dual-band fly-back spectral-spatial pulse design as a simple alternative to previously proposed numerical approaches. The design time was less than a second on a dual 2.8 GHz Quad-Core computer. We proposed a 2D pulse that can simultaneously reduce through-plane signal loss and suppress lipid signal. We also proposed a 4D pulse that can simultaneously reduce signal loss and B1+ inhomogeneity. The pulses were demonstrated to significantly reduce these artifacts in phantom and human brain gradient echo imaging experiments at 3T.

Both pulses performed comparably to the original numerical pulse design with regards to recovery of susceptibility induced signal loss. A value of $\alpha = -0.8 \mu\text{T/m/Hz}$ was determined to provide the largest visible amount of correction in all slices and all subjects. Although the previous study of the numerical design found a value of $-2.0 \mu\text{T/m/Hz}$ to work well, our own comparisons observed no large differences for values from ranging from -0.8 to $-2.0 \mu\text{T/m/Hz}$. The signal loss correction was in general very effective in all slices and all subjects. It was noticed, however, that the correction was not equally successful in all slices. Although the differences were not large, smaller magnitudes of α appeared to work better for the lower slices. We also observed small differences between the images acquired with the analytical and numerical 2D spectral-spatial pulses. One would expect the numerical design to provide a better slice profile because it uses an image domain approach as well as higher frequency resolution.

It is worth noting that the fly-back approach is not the only way to design the dual-band pulses. For example, one can design a 2D spectral-spatial pulse for each band using both the positive and negative z -gradient lobes and then add the two pulses together. This design appeared to work well in separate investigations, but did not offer any significant avenue for improved performance. Furthermore, this design will be more sensitive to timing and B0 errors than the fly-back design. An important limitation in the spectral-spatial pulse approach is that the length of the pulse limits the frequency resolution. Some brain regions have susceptibility induced signal loss at smaller frequency shifts such as 50 Hz. This would require a pulse on the order of 20 ms, which may be impractically long in duration. Further work is needed to measure the through-plane signal loss as a function of frequency to further optimize the pulse parameters.

The 4D design showed reduced B1+ inhomogeneity from the body coil. This was observed in all subjects and all slices. The inhomogeneity correction results were most clearly seen in the phantom images where the B1+ inhomogeneity was larger. In the human brains, the degree of B1+ inhomogeneity was smaller and required less compensation. We found a value of $\varepsilon = 0.5$ worked adequately for most slices and subjects. Possibly larger values of ε could be used to correct larger degrees of B1+ inhomogeneity at higher fields such as 7T, however, it is not clear if the geometry would satisfy the simple quadratic model. It is possible to further optimize the shape of the correction by including unique amplitude and phase terms on each of the k_x - k_y points. This may be helpful at higher fields or in cases where the subject is not centered in the body coil. There are also other possible 4D designs

that one could use, including utilizing fast- k_z pulse for both bands. This may be helpful if the off resonant regions suffer from B1+ inhomogeneity. However, the separation between aliasing lobes will also become smaller (approximately 227 Hz) due to the increased sub-pulse length and the frequency requirements may not be met.

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FIG 1.

Diagrams showing the real and imaginary parts of the RF waveforms as well as the x , y , and z gradient waveforms for the (a) 2D and (b) 4D spectral-spatial pulses designed with the analytical dual-band approach. The dashed box indicates a single five-rung fast- k_z sub-pulse. These pulses were used in the simulations and *in-vivo* experiments.

FIG 2.

Bloch equation simulations of the magnitude and phase of the magnetization produced by the pulses shown in Fig. 1. Figure (a) is the magnetization produced by the 2D pulse and (b) is from the 4D pulse. Each figure shows the x - y magnetization as functions of the slice-select (z) and the frequency (f) directions. Note that frequency bands at 0 Hz and 125 Hz can be seen. The magnetization at 125 Hz has a linear phase along the slice-select direction to compensate for the through-plane susceptibility gradient.

FIG 3. Images of a single oil phantom bundled with three water phantoms acquired at a 10 ms TE. (a) Control image excited with the standard pulse showing both water and lipid phantoms. (b) Image excited with the standard pulse and using fat saturation. (c) Image excited with the 2D spectral-spatial pulse. The images acquired using fat saturation and the 2D spectral-spatial pulse show reduced lipid magnitude compared to the control image. The bar graph displays the percent of image magnitude in each of the ROI's drawn on the images.

FIG 4.

(a) Images of a spherical water phantom acquired at a 10 ms TE. The upper left image was acquired with the standard pulse. Note the overly bright center characteristic of B1+ inhomogeneity from a volume coil. The upper right, lower left, and lower right are images acquired with the 4D spectral-spatial pulse using increasing degrees of B1+ inhomogeneity compensation $\varepsilon = 0.5, 1.0$, and 1.5 , respectively. (b) Plots of the magnitude profile from left to right across each phantom image.

FIG 5.

Gradient echo brain images acquired from a human subject at 3T. (a) Control images excited with the standard pulse using (a) TE = 10 ms and (b) TE = 30 ms. Note the signal loss susceptibility artifacts in (b). Images excited with the (c) 2D and (d) 4D spectral-spatial pulses as well as with the (e) numerical design using TE = 30 ms. All show a comparable degree of signal loss recovery. The images in (d) have improved B1+ homogeneity as seen by reduced brightness in the center.