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Spectral Decomposition of Susceptibility Artifacts for Spectral-Spatial RF Pulse Design

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Abstract

Susceptibility induced signal loss is a limitation in gradient echo functional MRI. The through-plane artifact in axial slices is particularly problematic due to the inferior position of air cavities in the brain. Spectral-spatial RF pulses have recently been shown to reduce signal loss in a single excitation. The pulses were successfully demonstrated assuming a linear relationship between susceptibility gradient and frequency, however, the exact frequency and spatial distribution of the susceptibility gradient in the brain is unknown. We present a spiral spectroscopic imaging sequence with a time-shifted RF pulse that can spectrally decompose the through-plane susceptibility gradient for spectral-spatial RF pulse design. Maps of the through-plane susceptibility gradient as a function of frequency were generated for the human brain at 3T. We found that the linear relationship holds well for the whole brain with an optimal slope of $-1.0\mu\text{T}/\text{m/Hz}$.

Keywords

Susceptibility artifacts; spectral-spatial RF pulses; spectroscopic imaging; BOLD fMRI

INTRODUCTION

High field MRI has several advantages including greater signal-to-noise ratio (1) and improved contrast for susceptibility weighted methods such as functional MRI (fMRI) (2). Although high fields offer great promise they also suffer from limitations due to increased artifacts including susceptibility artifacts that appear as distortions or as signal loss in the images. Susceptibility artifacts 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. The signal loss artifact leaves many regions that are key to our basic understanding of the brain and disease, such as the inferior frontal cortex, challenging to image. Therefore, there is a need for a better understanding of susceptibility artifacts and the development of improved methodology that reduce these artifacts.

Numerous methods have been proposed to mitigate the susceptibility induced signal loss artifact. These include z-shim methods (3–5), slice averaging (6), shims (7,8), multi-echoes (9–11), parallel transmission (12), and tailored RF pulses (13–15). Methods that reduce signal loss in a single excitation are of particular importance because they do not increase scan time. Spectral-spatial RF pulses have recently been shown to be capable of reducing

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signal loss in a single excitation with only one pulse (16–18). This method assumes that regions with signal loss are also off resonance and only compensate those frequencies while leaving on-resonant spins uncompensated.

Optimization of the above techniques requires an accurate mapping of the through-plane susceptibility gradient as a function of position. Furthermore, the spectral-spatial pulse method also needs a map of off-resonance frequency as a function of position. The original work of Yip *et al.* on the spectral-spatial pulse method found that assuming linear relationship between the through-plane susceptibility gradient and frequency $G_s(f) = \alpha f$ with $\alpha = -2.0 \mu\text{T/m/Hz}$ worked well. Although the authors experimentally confirmed this using a B0 map with contiguous 1 mm thick slices, this was not the main thrust of the paper and the precise spatial and frequency distribution of $G_s(f)$ over the whole brain in several subjects was not investigated.

We present a spiral spectroscopic imaging sequence with an incrementally shifted RF pulse that can be used to find $G_s(f)$. This method is potentially advantageous over a B0 map because it can directly image signal loss as a function of frequency for each pixel. This method was applied to understanding the spatial and frequency dependence of $G_s(f)$ for optimal “single-pulse” spectral-spatial pulse design; however, it may be of general interest for other applications that require knowledge of $G_s(f)$. The technique was used to acquire images of the through-plane signal loss as a function of frequency in the human brain at 3T. Fits to the image signal intensity were then performed to estimate $G_s(f)$ at different pixels and slice locations. The linear approximation was observed to hold well and a fit was performed for subjects and a value of $\alpha = -1.0 \mu\text{T/m/Hz}$ was determined. Spectral-spatial pulses were then designed with values of $\alpha = -1.0$ and $-2.0 \mu\text{T/m/Hz}$ and compared to a standard *sinc* slice-select pulse with an Echo Planar Imaging (EPI) fMRI sequence using a 30 ms TE at 3T. Both spectral-spatial pulses were found to reduce the signal loss artifact in all slices and all subjects with the $\alpha = -1.0 \mu\text{T/m/Hz}$ pulse recovering significantly more pixels than the $\alpha = -2.0 \mu\text{T/m/Hz}$ pulse. In no cases did we observe that the spectral-spatial pulses introduced additional signal loss.

THEORY

SPECTRAL DECOMPOSITION OF THE SIGNAL LOSS ARTIFACT

The magnetization $M(\mathbf{r})$ in a gradient echo axial slice at TE in the presence of a spatially dependent through-plane susceptibility gradient $G_s(x, y)$ can be written as

$$M(\mathbf{r}) = M_0(x, y)P(z)e^{i\gamma G_s(x, y)z\text{TE}} \text{ where } P(z) = \text{rect}(z/\Delta z). \quad [1]$$

Here $M_0(x, y)$ is the equilibrium magnetization and $P(z)$ is the slice profile with thickness Δz . The susceptibility gradient $G_s(x, y)$ induces a spatially dependent signal loss on the signal $s(x, y)$ in the imaged slice that has a *sinc* dependence:

$$\begin{aligned} s(x, y) &\propto M_0(x, y) \int_{-\infty}^{\infty} \text{rect}(z/\Delta z) e^{-i\gamma G_s(x, y)z\text{TE}} dz \\ &= M_0(x, y) \text{sinc}[\gamma G_s(x, y)\Delta z\text{TE}/2]. \end{aligned} \quad [2]$$

Signal loss at a pixel can be compensated by adjusting the area of the slice-select re-winder or by time shifting the RF pulse Δt with respect to the slice-select gradient G_z . These approaches are often termed *z-shim*. The signal equation can be modified to include the effect of a *z-shim*:

$$s(x, y) \propto M_0(x, y) \text{sinc} [\gamma (G_s(x, y) - \Delta G) \text{TE} \Delta z / 2] \text{ where } \Delta G \equiv G_z \Delta t / \text{TE}. \quad [3]$$

The signal loss is removed when $\Delta G = G_s(x, y)$. The drawback with the z -shim approach is that several increments of ΔG are needed to completely compensate for the spatial dependence of $G_s(x, y)$. This is impractical for many applications including fMRI due to the time penalty. It is therefore desirable to create a “single-shot” z -shim that can apply several shims simultaneously.

SPECTRAL-SPATIAL PULSES FOR REDUCED SIGNAL LOSS ARTIFACT

Spectral-spatial pulses can excite magnetization with both spatial and frequency specificity. The small tip-angle approximation provides an intuitive method for spectral-spatial pulse design using the concept of excitation k -space (19). In this picture, one can think of the excited slice profile as being proportional to the Fourier transform of the RF pulse $B_1(t)$ of length T .

$$P(z, f) \propto \int_0^T B_1(t) e^{ik_z(t)z + i2\pi f(t-T)} dt \text{ where } k_z(t) = -\gamma \int_t^T G_z(s) ds. \quad [4]$$

The slice profile $P(z, f)$ has been generalized to be frequency dependent. A spectral-spatial pulse is typically designed using a train of slice-select sub-pulses of length T_z where the gradient G_z is a series of trapezoids with alternating polarity (20). The periodicity of the sub-pulses produces a temporal sampling where the separation between spectral aliasing bands is given by $1/T_z$.

The frequency selectivity of a spectral-spatial pulse can be used to create a single-shot z -shim by making the assumption that regions with signal loss are also off resonance. The pulse design therefore requires a mapping of the spatial variation of the through-plane susceptibility gradient to a frequency variation. Yip *et al.* have demonstrated that linear frequency dependence was a good approximation:

$$G_s(x, y) \Rightarrow G_s(f) = \alpha f. \quad [5]$$

This work found in a single slice measurement that the slope α was on the order of $\alpha = -2.0 \mu\text{T/m/Hz}$. The spectral-spatial pulses can then be designed using the small tip angle approximation where the desired profile has a compensatory through-plane phase term to cancel the susceptibility gradient:

$$P(z, f) = \text{rect}(z/\Delta z) e^{-i\gamma \alpha f z \text{TE}}. \quad [6]$$

In the actual design process, the small tip angle equation is written as a matrix equation and $B_1(t)$ is solved using a least squares minimization approach such as the conjugate gradient method.

SPECTRAL DECOMPOSITION OF THE THROUGH-PLANE SUSCEPTIBILITY GRADIENT

The aims of this work are to examine the validity of the linear assumption given by Eqn. [5] as well as determine the optimal value of α and its spatial dependence in the human brain at 3T. These can be accomplished by incrementally time-shifting the RF pulse each TR in a spectroscopic imaging sequence that excites the water peak. The pixel signal intensity will now be a function of both frequency and ΔG .

$$s(x, y, f, \Delta G) \propto M_0(x, y, f) \text{sinc} [\gamma(\alpha f - \Delta G) \text{TE} \Delta z / 2]. \quad [7]$$

The frequency dependent magnetization is given by $M_0(x, y, f)$. The peak pixel signal intensity at a given frequency will occur when $\Delta G = \alpha f$.

METHODS

A spiral spectroscopic imaging sequence was constructed for spectral decomposition by taking a standard single echo spiral sequence and adding the capability of multiple echoes. The parameters were ten 64×64 5 mm axial slices, four interleaves, 22 cm FOV, 2000 ms TR, a train of 32 spiral echo readouts, and 32 pulse shifts. The length of each spiral including re-winders was 4.1 ms (2.5 ms spiral readout), which produced a spectral bandwidth of 256 Hz and a sampling interval of 7.7 Hz over a range of echo times from 30 to 130 ms. The spiral design used a slew rate of 150 T/m/s and a peak gradient of 30 mT/m. Each RF pulse shift step Δt was set such that the through-plane gradient was measured in increments of $\Delta G = 13.0 \mu\text{T/m}$. A standard fat saturation pulse was used such that only brain water content was imaged. The total scan time was 4:16. Figure 1 shows a diagram of the spiral spectroscopic imaging sequence with a shifted RF pulse.

Spectral-spatial pulses were designed using Matlab (The Mathworks, Natick, MA) code obtained from the “Image Reconstruction Toolbox” (J. A. Fessler) available online at <http://www.eecs.umich.edu/~fessler> running on a dual 2.8 GHz Quad-Core Macintosh Pro (Apple, Cupertino, CA) workstation. These are the pulses proposed in the original paper by Yip *et al.* (16) which should be consulted for a complete description of the method. The design had seven alternating z -gradient trapezoids using a slew rate of 150 T/m/s and a peak gradient of 30 mT/m. The desired excitation profile $P(z, f)$ was a *rect* function along the z -direction with a 5 mm width, 20 cm FOV, and a resolution of 1.7 mm. The frequency bandwidth was 500 Hz with a resolution of 1.7 Hz. Two spectral-spatial pulses were constructed with through-plane compensation terms of $\alpha = -1.0$ and $-2.0 \mu\text{T/m/Hz}$. The total length of the pulse was approximately 12 ms including a small z -gradient re-winder. Figure 2 shows a diagram of the spectral-spatial pulse with $\alpha = -2.0 \mu\text{T/m/Hz}$. The spectral-spatial pulses were then inserted in a standard fMRI Echo Planar Imaging (EPI) sequence with parameters of eight 64×64 5 mm axial slices, 22 cm FOV, 2000 ms TR, 30 ms TE. A standard fat saturation pulse was used.

All scanning was performed on a Siemens (Siemens Healthcare, Erlangen, Germany) TIM Trio 3T whole body scanner with a body coil transmitter and a 12-channel array receiver. All human subjects underwent informed consent using protocols approved by the University of Hawaii and Queens Medical Center Internal Review Board. Human brain images were obtained on six healthy adults using the spiral spectroscopic imaging sequence. Four human subjects were scanned in separate sessions using the EPI sequence run three times: once with a standard *sinc* pulse and then with each spectral-spatial pulse. The three EPI scans were run only once in two of the subjects and three times in the other two (with the head repositioned between each group of three) generating eight EPI data sets for all three pulses. In all scanning sessions a three-plane localizer was used to align the slices such that the middle slice was on the bottom edge of the frontal lobe.

The spectroscopic imaging data was processed offline using custom Matlab programs running on the Macintosh Pro workstation. First, for each slice, each image in the echo train was reconstructed using a standard spiral gridding reconstruction. A Fourier transform was then applied across echoes for each pixel. This generated the four-dimensional slice data $s(x, y, f, \Delta G)$ given by Eq. [7]. The x - y dimension was resampled to 16×16. Fits to $s(x, y, f,$

ΔG) using a *sinc* function at each frequency along the shift direction were then performed for every pixel and slice:

$$\psi_n(f, \Delta G) = A(f) \text{sinc} [(\Delta G - G_s(f)) / W(f)] \quad [8]$$

The three fit parameters $A(f)$, $G_s(f)$, and $W(f)$ represent the amplitude, through-plane susceptibility gradient, and width, respectively. The width term was required to model a broadening of the *sinc* modulation reflecting a distribution of through-plane susceptibility gradients. The index n represents each pixel and slice in a subject ($16 \times 16 \times 10$ values). The spectral-spatial pulse design parameter α could then be determined using linear fits to $G_s(f)$.

RESULTS

Figure 3 shows examples of a spectrally decomposed slice from a subject acquired using the spectroscopic imaging sequence. Figure 3(a) shows the slice displayed as individual images in x - y distributed over ΔG and frequency. Note that the regions where susceptibility artifacts are typically observed such as above the sinus and ear canals are shifted in both the ΔG and frequency directions. Figure 3(b) shows an example of fits from Eqn. [8] for the slice shown in (a) at 105 Hz. A mask was used to exclude pixels that were below a threshold of 10% of the maximum of $A(f)$. Note the pixels are shifted by approximately 75 $\mu\text{T/m}$ and signal loss would be expected. Figure 3(c) shows images of $G_s(f)$ for all tens slices in the subject determined by fitting all frequencies and slices with Eq. [8] (the slice shown in (a) is slice 5). The pixel magnitudes are windowed between -100 (black) and 100 (white) $\mu\text{T/m}$. Visual inspection of (c) indicates that the linear relationship $G_s(f) = \alpha f$ appears to hold for the majority of brain, particularly at the higher frequencies. However, at lower frequencies, there are slices that contain both positive and negative susceptibility gradients.

Figure 4 shows a scatter plot of $G_s(f)$ for all six subjects obtained from susceptibility gradient maps as shown in Fig. 3(c). The gray points are the gradient values for all pixels and slices and the black circles are the average gradient at each frequency. The error bars are the standard deviation for all pixel and slices at that frequency with a mean equal to 20 $\mu\text{T/m}$ across frequency. The linear relationship appears to hold throughout the brain in all subjects within a range of through-plane gradients approximately between -150 and $100 \mu\text{T/m}$ and frequencies approximately between -75 and 150 Hz. The average values indicate that there is a slight upturn between approximately ± 25 Hz, which is consistent with the presence of both positive and negative gradients in this frequency range in Fig. 3(c). The dashed line shows a linear fit to the average gradient values for frequencies greater than 25 Hz that yielded a value of $\alpha = -1.0 \mu\text{T/m/Hz}$. The standard deviation of α between subjects was $0.1 \mu\text{T/m/Hz}$.

Figure 5 shows the gradient echo EPI slices in a sample subject acquired at 3T with a 30 ms TE. The row (a) shows slices that were acquired using a standard *sinc* slice-select pulse. Rows (b) and (c) show the same slice acquired with spectral-spatial pulses designed for $\alpha = -1.0$ and $-2.0 \mu\text{T/m/Hz}$ respectively. The slices acquired with the spectral-spatial pulses in (b) and (c) show recovered image intensity in regions with susceptibility artifacts in (a). Rows (d) and (e) show (b)-(a) and (c)-(a) displayed as masks for difference values larger than 50% of the mean of each respective image in row (a). Visual inspection shows that the $\alpha = -1.0 \mu\text{T/m/Hz}$ pulse gives more regions of improved signal recovery compared to the $\alpha = -2.0 \mu\text{T/m/Hz}$ pulse. Difference masks in all slices were summed for all eight runs to find the total number of recovered pixels for each spectral-spatial pulse. Table 1 lists the mean number of recovered pixels in the all slices for each of the eight runs and the p-values obtained by a paired t-test between slices. All runs show significant improvements with the

$\alpha = -1.0 \mu\text{T/m/Hz}$ pulse compared to the $\alpha = -2.0 \mu\text{T/m/Hz}$ pulse. A second paired t-test across runs yielded a p-value of 8.1×10^{-5} . The differences (a)–(b) and (a)–(c) were also taken and no loss of pixel intensity was observed by using either spectral-spatial pulse.

DISCUSSION AND CONCLUSIONS

This paper presented the use of spectroscopic imaging with an incrementally shifted RF pulse to decompose the through-plane susceptibility gradient as a function of position and frequency. This technique was applied to the problem of spectral-spatial RF pulse design for susceptibility artifact reduction in gradient echo imaging applications such as fMRI. However there may be other applications for which this method may be useful. We found that the optimal linear term was $\alpha = -1.0 \mu\text{T/m/Hz}$ for the whole brain at 3T. Spectral spatial pulses were designed using this value as well as $\alpha = -2.0 \mu\text{T/m/Hz}$, which was the value utilized in the original paper demonstrating spectral-spatial pulses for signal loss reduction (16). Both pulses provided whole brain signal loss recovery using EPI scans with the $\alpha = -1.0 \mu\text{T/m/Hz}$ pulse giving significantly more recovered pixels than the $\alpha = -2.0 \mu\text{T/m/Hz}$ pulse. The value of $\alpha = -1.0 \mu\text{T/m/Hz}$ was also in good agreement with empirically derived α values in our previously published work on using spectral spatial pulses for simultaneous B1+ and signal loss mitigation (17,18).

We found the relationship between susceptibility gradient and frequency to be well approximated as linear, although there were some deviations. These can be seen in Figs. 3(c) and 4 where both positive and negative susceptibility gradients were seen close to center frequency (± 25 Hz). These positive shifts could theoretically produce signal loss, however, the range of gradients is small ($\pm 20 \mu\text{T/m}$ in one standard deviation) and were not observed to create artifact. Furthermore, in no subjects did we observe that signal loss was introduced using either spectral-spatial pulse compared to the standard pulse. There are limitations with our measurement including signal loss from in-plane susceptibility gradients and blurring. These were not noticed to be a major confound. In general, the linear approximation and a single spectral-spatial pulse appear to be adequate for applications such as fMRI at 3T. Further investigations are still needed to quantify BOLD signal recovery using spectral-spatial pulse methods.

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FIG 1.

This figure shows a diagram of a spiral spectroscopic imaging pulse sequence with shifted RF pulse. The RF pulse is shifted a multiple of Δt each TR to spectrally decompose the *sinc* modulation produced by the through-plane magnetic susceptibility gradient as a function of position and frequency. The rows from top to bottom are the RF waveforms, the *z*-gradient, *x*-gradient, and *y*-gradient.

FIG 2.

This figure shows a diagram of spectral-spatial RF pulse that compensates signal loss assuming that the through-plane magnetic susceptibility gradient is a linear function of frequency with slope $-2.0 \mu\text{T/m/Hz}$. The top row shows the z -gradient and the bottom row shows the real and imaginary parts of the RF pulse waveform as solid and dotted lines. The oscillatory property of the z -gradient provides frequency selectivity.

FIG 3.

Image (a) shows an example of a spectrally decomposed brain slice from a human subject acquired at 3T with the sequence shown in Fig. 1. The slice is displayed as individual images distributed over the through-plane gradient shift and frequency. The plots in (b) show *sinc* fits to the pixel intensities (above 10% of the maximum) at 105 Hz in (a). Images of $G_s(f)$ (windowed between $\pm 100 \mu\text{T/m/Hz}$) for all ten slices in the same subject are shown in (c). Slice 5 is the slice shown in (a).

FIG 4.

This plot shows the through-plane susceptibility gradient as a function of frequency for all subjects. The gray points represent the gradient values for all pixels and slices for all subjects. The circles are the average susceptibility gradient at each frequency including error bars reflecting the standard deviation. The dashed line is a linear fit to the average values above 25 Hz giving a slope equal to $-1.0 \mu\text{T/m/Hz}$.

FIG 5.

This figure shows gradient echo EPI slices in a subject at 3T using a TE of 30 ms. Row (a) shows slices that were acquired using a standard slice-select pulse and rows (b) and (c) show slices acquired using spectral-spatial pulses with $\alpha = -1.0$ and $-2.0 \mu\text{T/m/Hz}$. Rows (d) and (e) show binary masks of where differences between rows (b) and (c) from (a) are observed.

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Table 1

Mean number of recovered pixels in the all slices per subject from EPI scans using spectral-spatial pulses. The p-values were determined by a paired t-test across slices in each run. A second paired t-test across runs yielded a p-value of 8.1×10^{-5} .

	Run 1	Run 2	Run 3	Run 4	Run 5	Run 6	Run 7	Run 8
$\alpha = -1.0 \mu\text{T/m}$	161	135	119	169	250	226	263	235
$\alpha = -2.0 \mu\text{T/m}$	85	77	62	87	190	104	134	119
p-value	0.0063	0.0006	0.0013	0.0002	0.0099	0.0037	0.0015	0.0317