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Parallel Excitation in the Human Brain at 9.4 T counteracting K-Space Errors with RF Pulse Design

Xiaoping Wu¹, J. Thomas Vaughan¹, Kâmil Uğurbil¹, and Pierre-François Van de Moortele¹

¹ Center for Magnetic Resonance Research, Department of Radiology, University of Minnesota Medical School, Minneapolis, Minnesota, USA

Abstract

Multidimensional spatially selective RF pulses have been proposed as a method to mitigate transmit B1 inhomogeneity in MR experiments. These RF pulses, however, have been considered impractical for many years because they typically require very long RF pulse durations. The recent development of parallel excitation techniques makes it possible to design multidimensional RF pulses that are short enough for use in actual experiments. However, hardware and experimental imperfections can still severely alter the excitation patterns obtained with these accelerated pulses. In this note, we report at 9.4 T on a human eight-channel transmit system, substantial improvements in 2D excitation pattern accuracy obtained when measuring k-space trajectories prior to parallel transmit RF pulse design (acceleration $\times 4$). Excitation patterns based on numerical simulations closely reproducing the experimental conditions were in good agreement with the experimental results.

Introduction

Multidimensional spatially selective RF pulses (1,2) have been proposed in a variety of MR applications, including arbitrarily defined excitation volumes (3) and transmit B1 (B1+) inhomogeneity mitigation at high field (4). However, due to limited gradient system capability, these RF pulses typically require a long duration (several tens of milliseconds) which makes them impractical due to spin transverse relaxation, B0 inhomogeneity and cumulative gradient errors. These problems are even more pronounced at very high field due to larger B0 non-uniformity and shorter transverse relaxation time constants. It has been shown that these limitations can be overcome using parallel excitation techniques (5-12) allowing for short, accelerated selective RF pulses. These accelerated RF pulses are especially promising at very high field where B1+ profiles are significantly heterogeneous (13). However, even with such shortened RF pulses significant degradation of excitation patterns can occur, especially in the case of gradient system imperfections.

Previous studies have shown with conventional single-channel transmit (Tx) systems that using the actual, measured gradient waveforms (or k-space trajectory) for 2D RF pulse calculation can effectively reduce gradient waveform induced excitation pattern errors (14,15). This has also been demonstrated in parallel excitation on an animal system with four Tx channels (16). To date, parallel excitation with human subjects has been reported at field strengths up to 7 T using 2D (17) and 3D (12) RF pulses, but to the best of our

Address for correspondence: Pierre-François Van de Moortele, M.D., Ph.D. Center for Magnetic Resonance Research University of Minnesota Medical School 2021 6th Street S.E. Minneapolis, MN 55455 Phone: (612) 625 6526 (office) / 626 2001 (front desk) Fax: (612) 626 2004 pfvdm@cmrr.umn.edu.

knowledge this has not been reported at higher field. Furthermore, we are not aware of k-space trajectory calibrations in humans for multi-Tx RF pulse design.

In this note we report for the first time successful parallel excitation results in human brain at 9.4 T with eight Tx channels. For this implementation it was necessary to measure the actual gradient waveforms in addition to ΔB_0 maps before designing 2D selective RF pulses (acceleration $\times 4$) to address severe distortion and blurring initially observed in excitation patterns. Experimental results were highly consistent with numeric simulations for both in-vivo and phantom data.

Materials and Methods

All experiments were performed on a 9.4 T, 65-cm inner diameter bore human scanner (Magnex, UK), equipped with a head gradient coil capable of a maximum gradient strength of 50 mT/m and maximum slew rate of 166 T/m/s. The scanner was driven by a console with eight independent Tx RF channels (DirectDrive, Varian, USA), each powered with a 500W RF amplifier (CPC, Brentwood, NY, USA). To ensure compliance with the FDA guidelines on specific absorption rate¹, the RF output power of each of the eight RF amplifiers was continuously monitored with an in-house-built multi-channel monitoring unit. Each RF amplifier was equipped with a directional coupler providing a 50dB attenuated version of the output waveform whose power envelope was continuously sampled by a calibrated ADC board. The system was maintaining a 10-second and a 10-minute moving average and would immediately disable all RF amplifiers and terminate the MR pulse sequence if the mean power exceeded a predefined wattage threshold. Thresholds were conservatively determined against the worst case scenario assuming all RF coil electric fields to add constructively in the sample. Additionally, only small flip angle RF excitation with low level RF output power was utilized for in-vivo experiments. MR signals were acquired on eight independent receive (Rx) channels with a home built digital receiver system based on Echotek boards (Huntsville, AL, USA). All computations, including RF pulse design and Bloch simulations, were conducted in Matlab (MathWorks Inc.). Three healthy volunteers were recruited for this study, which was approved by the local institutional review board. All subjects provided written consent. Representative results are shown for one subject.

Parallel Excitation Measurements

All parallel excitation experiments were performed with an elliptical eight-channel transceiver stripline array, similar to the coil described in (18). Each channel of the stripline array was used for both transmission and reception. A 16 cm diameter spherical saline phantom (NaCl 99 mM) doped with copper sulfate ($T_1 \sim 200$ ms) was used for preliminary measurements and demonstration of 2D parallel transmission.

Analytical Gradient Waveform Design

2D gradient waveforms (G_x and G_y), corresponding to a spiral excitation k-space trajectory, were designed for accelerated spatially selective excitation. Slew rate-limited gradient waveforms were first calculated based on G. Glover's algorithm (19) at the maximum available slew rate (166 T/m/s) to generate a spiral-in excitation k-space trajectory that radially undersampled the k-space by a factor of four (Fig. 1a and b). The fully sampled k-space corresponded to a field of excitation (FOX) of 16×16 cm² with a spatial resolution of 5 mm. Gradient ramps were appended at the beginning of these initially calculated gradient

¹U. S. Food and Drug Administration (2003) Criteria for Significant Risk Investigations of Magnetic Resonance Diagnostic Devices. <http://www.fda.gov/MedicalDevices/DeviceRegulationandGuidance/GuidanceDocuments/ucm072686.htm>

waveforms as they typically start with nonzero values on at least one axis. The resulting gradient waveforms had a total duration of 2.34 ms with a temporal resolution of 4 μ s.

Actual Gradient Waveform Measurement

Even with application of system pre-emphasis for eddy current compensation, significant distortion and blurring were found to be present in excitation patterns with 2D RF pulses based on spiral gradient waveforms calculated with the ideal analytical definition mentioned above. Thus, actual Gx and Gy were measured in separate experiments to identify potential gradient waveform deviations, using a self-encoding gradient method as previously described in (14). This method relies on a modified gradient echo (GE) pulse sequence where the Cartesian encoding steps along the phase direction are replaced with a variable self-encoding predephasing gradient followed by the gradient waveform to be measured used as a readout gradient during which the MR signals are sampled. Note that, as in (14), we measured Gx and Gy independently assuming no significant cross terms between the two. To limit induced eddy currents while providing whole excitation k-space coverage, the trapezoid self-encoding gradient had a long duration (11 ms) with limited maximum gradient strength. To improve automatic peak signal finding, homogeneous Tx and Rx B1 distributions were obtained using a small doped water tube (1 cm in diameter, 170 mM NaCl and 0.3 mM NiCl₂, T1 ~ 1300 ms) set in the center of a one-channel ring shaped Tx/Rx RF coil of 15 cm in diameter. Slice selective acquisitions were obtained in an axial plane, in this single-channel Tx/Rx setup, with slice thickness = 5 mm, TR/TE = 100/14.4 ms, bandwidth = 250 kHz, and 96 self-encoding steps. Similar to (14), *reception* k-space trajectories (Kx and Ky) were extracted from these data by locating the maximum signal values along the self-encode dimension. The peak position at each time point was determined by thresholding the magnitude signal to only contain its main lobe and fitting this lobe with a second-degree polynomial to find the position of its maximum. To minimize noise impact, ten data acquisitions were averaged before extracting k-space trajectories, yielding a total acquisition time of 6 min 24 s for both Gx and Gy characterizations. These trajectories were smoothed segment by segment, before calculating their time derivative to obtain the actual gradient waveforms. In addition, *baseline* k-space trajectories were obtained by running the same acquisitions (for Gx and for Gy), but *without* the application of the spiral gradient waveforms. These *baseline* trajectories, which are reflecting B0 inhomogeneities but not the actual gradient waveforms, were subtracted from the previously measured trajectories just before final waveform calculations.

Transmit B1, Receive B1 and Δ B0 Field Mapping

In the multi-transmit setup, complex valued B1+ maps were obtained for each individual channel with a fast multi-channel B1+ mapping method introduced in (20). With this method, eight *relative* B1+ maps (magnitude and phase), derived from a series of GE images collected in the small flip angle regime with one channel transmitting at a time, are combined with one *absolute* magnitude B1+ map, obtained in the large flip angle regime with all channels transmitting together (20). The eight small flip angle GE images (one per transmit channel) were acquired with TR/TE = 200/4 ms, slice thickness = 5 mm, FOV = 20 $\times$ 20 cm², matrix size = 128 $\times$ 64, and acquisition time = 2 min 4 s. The large flip angle map was obtained using the Actual Flip Angle (AFI) method introduced by V.L.Yarnyhk (21) with TR1/TR2/TE = 28/140/4.6 ms, FOV = 20 $\times$ 20 $\times$ 7.5 cm³, matrix size = 128 $\times$ 64 $\times$ 15, and acquisition time = 2 min 40 s. Note that, to avoid areas with very low B1+ due to B1+ destructive interferences between Tx channels (22) when arbitrarily using the native phase of each RF channel without modification, RF phase B1 shimming was first calculated based on the eight relative B1+ maps as described in (23,24) so that only one large flip angle AFI acquisition was needed (Fig. 2). The total acquisition time for B1+ estimations for the eight

channels was 4 min 24 s. Fig. 3 shows typical B1+ maps for both the phantom and human brain.

In addition to B1+ mapping, an estimation of the product (Rx B1 profile $\times$ proton density) was derived from GE images which were acquired using the same RF phase B1 shimming as for AFI based B1+ maps with about 15° flip angles in the center of the slice. Given the relatively short TR (150 ms) some degree of T1 contrast bias is expected in the product estimation, especially in the brain because of long T1 constant in tissues at 9.4 T (Fig. 2c).

ΔB_0 maps were calculated from two GE images acquired at different echo times (25) (TE1/TE2/TR = 8/9/60 ms, FOV = $20 \times 20 \text{ cm}^2$, matrix size = 128×64), and were incorporated into RF pulse design, to minimize deteriorations of excitation patterns due to off-resonance effects(7-9).

RF Pulse Design

In this study, parallel transmit pulses were designed in the small tip angle regime using Grissom's method (8). The optimization problem was formulated to minimize the residual error $\|A\mathbf{b} - \mathbf{m}\|_{(\mathbf{S}, \mathbf{T})}$. Here, matrix $\mathbf{A}$ is generated based on the excitation k-space trajectory, the ΔB_0 map and B1+ maps; vector $\mathbf{b}$ represents the eight complex RF pulse shapes; vector $\mathbf{m}$ represents the desired excitation pattern; the diagonal matrixes $\mathbf{S}$ and $\mathbf{T}$ define a spatial mask for the region of interest and a temporal mask for RF pulsing, respectively. The minimization problem was solved for $\mathbf{b}$ using conjugate gradient (CG) iterations along with the L-curve criterion (7). The desired excitation pattern was a rectangle with uniform flip angle and uniform phase distributions in a transverse plane, which was defined on a $20 \times 20 \text{ cm}^2$ region with a 5-mm resolution. Pixels without spins (located outside the phantom or outside the brain tissues) were excluded from RF pulse calculations. RF pulses were set to zero during the initial ramp of the gradient waveforms.

To investigate the impact of gradient error on excitation patterns, *uncorrected* and *corrected* parallel Tx RF pulses were calculated using the ideal and actual gradient waveforms, respectively. In all cases RF pulses were designed with ΔB_0 correction.

Parallel Transmit Experiment

All images of parallel excitation patterns were acquired with a modified 3D GE pulse sequence where the slice selective excitation is replaced by accelerated 2D spatially selective RF excitation. Relevant acquisition parameters were: TR/TE = 150/4 ms, FOV = $20 \times 20 \times 20 \text{ cm}^3$ and matrix size = $128 \times 64 \times 40$. Nominal flip angles were 17° for the phantom and 4° for human subjects. Maximum B1 magnitude during the pulse was 5.2 μT in human brain and 8 μT in phantom. The resulting images were divided by the sum of magnitudes of eight estimated proton density weighted Rx B1 profiles to mainly retain intensity variations due to RF excitation. In addition, FA distributions within the target rectangle were also mapped for the phantom, using AFI with TR1/TR2/TE = 28/140/4.6 ms, FOV = $20 \times 20 \times 20 \text{ cm}^3$ and matrix size = $128 \times 64 \times 40$. This FA mapping was not performed for in vivo experiments because the limited source of RF power together with a larger load of a human head than our phantom did not allow for large flip angles with the accelerated RF pulses. (For each 500W RF amplifier only about 200W is available at the RF coil).

Simulations

A Bloch simulator was developed in the spin domain (26) and was used for calculating the excitation patterns corresponding to *uncorrected* and *corrected* RF pulses. To reproduce as closely as possible the experimental conditions, Bloch simulations were based on the actual

gradient waveforms (rather than the ideal spiral counterpart) as well as on measured B_1+ and ΔB_0 maps. As will be seen, a time delay was observed between the actual gradient waveforms on X and Y axes. In order to distinguish between shearing distortions typically induced in excitation pattern by such delay and other pattern distortions, additional simulations using *uncorrected* RF pulses were obtained where the measured gradient waveforms were corrected for the time delay between X and Y axes. Excitation patterns were defined on a $20 \times 20 \text{ cm}^2$ FOX with a 128×64 matrix to match the experimental settings. All simulations assumed a uniform magnetization at equilibrium. Relaxation and diffusion terms were ignored.

Results and Discussion

The actual gradient waveforms were significantly different than their ideal counterparts, even though their overall pattern was similar (Fig. 1a). A delay of $12 \mu\text{s}$ was identified between the x and y gradient axes (y gradient events occurred earlier). In addition, actual gradient strengths were decreased by as much as 4% for G_x and 3% for G_y around the peak values of the waveforms. This resulted in significant k-space trajectory deviations (Fig. 1b), and the corresponding calculated RF pulses exhibited substantial differences, as illustrated when comparing the sum of magnitudes of uncorrected versus corrected RF pulses (Fig. 1c).

For both phantom and in-vivo experiments, the experimental excitation patterns obtained with RF pulses that had been designed based on the ideal gradient waveforms suffered strong blurring and geometric distortions (left column in Figs. 4 and 5). By contrast, using the actual gradient waveform for the RF pulse design provided significant improvement, resulting in excitation patterns matching very closely the target rectangle (right column in Figs 4 and 5).

Good agreement was observed between the experimentally obtained excitation patterns and the numeric simulations for both phantom and in vivo experiments (Figs. 4 and 5). Note, however, that these numeric simulations were strictly predicting excitation patterns, assuming homogeneous proton density and ignoring T_1 and T_2 relaxations. Because of long T_1 in brain tissues at 9.4 T, our in-vivo $R_x B_1$ estimation was partially contaminated with T_1 contrast, resulting in some visible brain structures (Fig 5b).

Consistent with others' recent work (27) our results show that obvious errors in excitation pattern (mostly shearing distortion and blurring) can be effectively addressed using appropriate pre-calibration of excitation k-space deviation in RF pulse design. Note that all experiments were obtained using the standard pre-emphasis of the system for eddy current compensation. The corresponding gradient errors primarily concerned synchronization between gradient axes, which is expected to translate into shearing distortion, as well as gradient strength fidelity. Our simulations with uncorrected RF pulses show that correcting for the delay between X and Y gradient waveforms compensated for the most part of the shearing distortion of the excitation patterns (Fig. 5d). However, strong blurring and fidelity errors still persisted after this correction, indicating that it was necessary to incorporate the whole measured gradient waveforms into the RF pulse design to restore excitation pattern fidelity. Note that, although we had satisfactory results using the peak finding approach for gradient measurements, other methods providing reduced acquisition time or increased SNR could be considered (28,29).

With large susceptibility induced B_0 inhomogeneity, one could expect additional difficulty to achieve accurate excitation patterns in parallel excitation at 9.4 T in humans. In this study gradient deviations were however the dominant source of phase errors. The limited impact of B_0 inhomogeneity, despite B_0 shim was not used, was likely due to the relatively short duration (2.34 ms) of the accelerated RF pulses.

In summary, we have demonstrated successful accelerated 2D selective RF excitations in human brain at 9.4 T, currently the highest magnetic field available for human studies, with eight Tx channels in the context of gradient errors and without using B0 shim.

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References

1. Pauly J, Nishimura D, Macovski A. A K-Space Analysis of Small-Tip-Angle Excitation. *Journal of Magnetic Resonance*. 1989; 81(1):43–56.
2. Hardy CJ, Cline HE. Spatial Localization in 2 Dimensions Using Nmr Designer Pulses. *Journal of Magnetic Resonance*. 1989; 82(3):647–654.
3. Bornert P, Schaffter T. Curved slice imaging. *Magnetic Resonance in Medicine*. 1996; 36(6):932–939. [PubMed: 8946359]
4. Saekho S, Yip CY, Noll DC, Boada FE, Stenger VA. Fast-kz three-dimensional tailored radiofrequency pulse for reduced B1 inhomogeneity. *Magnetic resonance in medicine : official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine*. 2006; 55(4):719–724. [PubMed: 16526012]
5. Katscher U, Bornert P, Leussler C, van den Brink JS. Transmit SENSE. *Magnetic Resonance in Medicine*. 2003; 49(1):144–150. [PubMed: 12509830]
6. Zhu Y. Parallel excitation with an array of transmit coils. *Magnetic Resonance in Medicine*. 2004; 51(4):775–784. [PubMed: 15065251]
7. Ullmann P, Junge S, Wick M, Seifert F, Ruhm W, Hennig J. Experimental analysis of parallel excitation using dedicated coil setups and simultaneous RF transmission on multiple channels. *Magn Reson Med*. 2005; 54(4):994–1001. [PubMed: 16155886]
8. Grissom W, Yip CY, Zhang Z, Stenger VA, Fessler JA, Noll DC. Spatial domain method for the design of RF pulses in multicoil parallel excitation. *Magnetic Resonance in Medicine*. 2006; 56(3):620–629. [PubMed: 16894579]
9. Setsompop K, Wald LL, Alagappan V, Gagoski B, Hebrank F, Fontius U, Schmitt F, Adalsteinsson E. Parallel RF transmission with eight channels at 3 Tesla. *Magn Reson Med*. 2006; 56(5):1163–1171. [PubMed: 17036289]
10. Xu D, King KF, Zhu Y, McKinnon GC, Liang ZP. A noniterative method to design large-tip-angle multidimensional spatially-selective radio frequency pulses for parallel transmission. *Magn Reson Med*. 2007; 58(2):326–334. [PubMed: 17654576]
11. Vernickel P, Roschmann P, Findekle C, Ludeke KM, Leussler C, Overweg J, Katscher U, Grasslin I, Schunemann K. Eight-channel transmit/receive body MRI coil at 3T. *Magn Reson Med*. 2007; 58(2):381–389. [PubMed: 17654592]
12. Setsompop K, Alagappan V, Gagoski B, Witzel T, Polimeni J, Potthast A, Hebrank F, Fontius U, Schmitt F, Wald LL, Adalsteinsson E. Slice-selective RF pulses for in vivo B1+ inhomogeneity mitigation at 7 tesla using parallel RF excitation with a 16-element coil. *Magn Reson Med*. 2008; 60(6):1422–1432. [PubMed: 19025908]
13. Collins CM, Yang QX, Wang JH, Zhang X, Liu H, Michaeli S, Zhu XH, Adriany G, Vaughan JT, Anderson P, Merkle H, Ugurbil K, Smith MB, Chen W. Different excitation and reception distributions with a single-loop transmit-receive surface coil near a head-sized spherical phantom at 300 MHz. *Magnetic Resonance in Medicine*. 2002; 47(5):1026–1028. [PubMed: 11979585]
14. Takahashi A, Peters T. Compensation of multi-dimensional selective excitation pulses using measured k-space trajectories. *Magnetic Resonance in Medicine*. 1995; 34(3):446–456. [PubMed: 7500885]

15. Bornert P, Aldefeld B. On spatially selective RF excitation and its analogy with spiral MR image acquisition. *Magma* (New York, NY). 1998; 7(3):166–178.
16. Ullmann, P.; Haas, M.; Hennel, F.; Wick, M.; Voiron, J.; Zaitsev, M.; Hennig, J.; Ruhm, W. Parallel Excitation Experiments Using Measured k-Space Trajectories for Pulse Calculation. Toronto: April. 2008 p. 1313
17. Setsompop, K.; Zelinski, AC.; Alagappan, V.; Nistler, J.; Hebrank, A.; Fontius, U.; Schmid, F.; Wald, LL.; Adalsteinsson, E. In vivo parallel RF excitation with B0 correction. Berlin, Germany: 2007. p. 6712007
18. Adriany, G.; Van de Moortele, PF.; Wiesinger, F.; Andersen, P.; Strupp, JP.; Zhang, X.; Snyder, C.; Chen, W.; Pruessmann, KP.; Boesiger, P.; Vaughan, JT.; Ugurbil, K. Transceive stripline arrays for ultra high field parallel imaging applications. Toronto, Canada: 2003. p. 474
19. Glover GH. Simple analytic spiral K-space algorithm. *Magnetic Resonance in Medicine*. 1999; 42(2):412–415. [PubMed: 10440968]
20. Van de Moortele, PF.; Snyder, C.; DelaBarre, L.; Adriany, G.; Vaughan, JT.; Ugurbil, K. Calibration Tools for RF Shim at Very High Field with Multiple Element RF Coils: from Ultra Fast Local Relative Phase to Absolute Magnitude B1+ Mapping. Berlin: 2007. p. 1676
21. Yarnykh VL. Actual flip-angle imaging in the pulsed steady state: a method for rapid three-dimensional mapping of the transmitted radiofrequency field. *Magnetic Resonance in Medicine*. 2007; 57(1):192–200. [PubMed: 17191242]
22. Van de Moortele PF, Akgun C, Adriany G, Moeller S, Ritter J, Collins CM, Smith MB, Vaughan JT, Ugurbil K. B1 destructive interferences and spatial phase patterns at 7 T with a head transceiver array coil. *Magnetic Resonance in Medicine*. 2005; 54(6):1503–1518. [PubMed: 16270333]
23. Van de Moortele, PF.; Snyder, C.; DelaBarre, L.; Akgun, C.; Wu, X.; Vaughan, T.; Ugurbil, K. Fast Mapping of Relative B1+ Phase in the Human Head at 9.4 Tesla with a 14 Channel Transceive Coil Array. Wuerzburg: 2006. p. 37Fev 16-18
24. Metzger G, Snyder C, Akgun C, Vaughan T, Ugurbil K, Van de Moortele PF. Local B1+ Shimming for Prostate Imaging with Transceiver Arrays at 7 Tesla Based on Subject Dependent Transmit Phase Measurements. *Magnetic Resonance in Medicine*. 2008; 59:396–409. [PubMed: 18228604]
25. Jezzard P, Balaban RS. Correction for geometric distortion in echo planar images from B0 field variations. *Magnetic Resonance in Medicine*. 1995; 34:65–73. [PubMed: 7674900]
26. Jaynes ET. Matrix Treatment of Nuclear Induction. *Phys Rev*. 1955; 98(4):1099–1105.
27. Schneider, J.; Haas, M.; Hennig, J.; Junge, S.; Ruhm, W.; Ullmann, P. Coping with Off-Resonance Effects and Gradient Imperfections in Parallel Transmission Experiments.. Proceedings 17th Scientific Meeting, International Society for Magnetic Resonance in Medicine; 2009. p. 172
28. Alley MT, Glover GH, Pelc NJ. Gradient characterization using a Fourier-transform technique. *Magn Reson Med*. 1998; 39(4):581–587. [PubMed: 9543420]
29. Duyn JH, Yang Y, Frank JA, van der Veen JW. Simple correction method for k-space trajectory deviations in MRI. *J Magn Reson*. 1998; 132(1):150–153. [PubMed: 9615415]

Fig. 1.

Gradient waveform and sum of magnitudes of RF pulses. (a) Nominal versus actually measured gradient waveforms. (b) Corresponding k-space trajectories. These slew rate limited gradients were designed for $\times 4$ acceleration (2.34 ms RF pulse duration). (c) Sum of magnitudes of the eight RF pulses based on nominal (uncorrected) and measured (corrected) trajectories. Note that no RF pulse was applied during the initial ramp of the gradient waveform. Also note the deviation of the actual k-space trajectories compared with their nominal counterparts, resulting in noticeable differences in shape between the sums of magnitudes of RF pulses.

Fig. 2.

Gradient Echo (GE) images and B1 profiles in phantom (top) and human brain (bottom) using a standard slice selective RF pulse. (a) Magnitude GE image (sum of magnitudes of the eight Rx channels). (b) Transmit B1 profile used for image acquisitions in (a). (c) Estimation of the product (receive B1 profile $\times$ proton density) derived from (a) and (b). Note visible brain structures in this estimation (bottom in c) because of proton density and because of some degree of T1 contrast due to long T1 values in tissue at 9.4 T. In each of these acquisitions all channels were transmitting together using a same standard slice selective RF pulse.

Fig. 3. Absolute magnitude (top) and relative phase (bottom) transmit B1 maps of each individual transmit channel in an axial slice in the phantom (a) and in the human brain (b).

Fig. 4.

Comparison of RF pulse performances in the phantom. (a) Rectangle target (black box) overlaid on an axial image. (b) Experimental excitation pattern for uncorrected (left) and corrected (right) parallel transmit RF pulses ($\times 4$ acceleration, 2.34 ms). Excitation patterns were imaged with a 3D GE sequence using the accelerated RF pulse with nominal flip angle = 17° . Estimated receive B1 profiles were removed from the experimental images to mainly retain image intensity variations due to transmit B1. Plots above images display intensity profiles along the red dashed line. Maps in colors show flip angles within the rectangle target, acquired with actual flip angle imaging. (c) Corresponding results obtained with Bloch simulations based on same acquisition parameters. Note the good agreement between experimental results and simulations.

Fig. 5.

Comparison of RF pulse performances in human brain. (a) Rectangle target (black box) overlaid on an axial image. (Note that gray scales were adjusted to help visualize residual brain structures within the rectangle target, resulting in excessive darkness in other areas of the brain). (b) Experimental excitation patterns for uncorrected (left) and corrected (right) parallel transmit RF pulses ($\times 4$ acceleration, 2.34 ms). Excitation patterns were imaged with a 3D GE sequence using the accelerated RF pulse with a nominal flip angle of 4° . Estimated receive B1 profiles were removed from experimental images, although residual T1 contrast was present due to long T1 in brain tissues. Plots above images display intensity profiles along the red dashed line. (c) Corresponding results obtained with Bloch simulations based

on same acquisition parameters. Maps in colors show simulated flip angles within the rectangle target. (d) Additional simulated excitation pattern using uncorrected pulses with the delay between gradient axes being compensated. Note significant improvement of excitation accuracy achieved using corrected pulses.