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Parallel Transmission RF Pulse Design for Eddy Current Correction at Ultra High Field

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Abstract

Multidimensional spatially selective RF pulses have been used in MRI applications such as B_1 and B_0 inhomogeneities mitigation. However, the long pulse duration has limited their practical applications. Recently, theoretical and experimental studies have shown that parallel transmission technique can effectively shorten pulse duration without sacrificing the quality of the excitation pattern. Nonetheless, parallel transmission with accelerated pulses can be severely impeded by hardware and/or system imperfections. One of such imperfections is the effect of the eddy current field. In this paper, we first show the effects of the eddy current field on the excitation pattern and then report RF pulse design method to correct eddy current fields caused by the RF coil and gradient system. Experimental results on a 7T human eight-channel transmit system show substantial improvements on excitation patterns with the use of eddy current correction. Moreover, the proposed model-based correction method not only demonstrates the comparable excitation pattern as the trajectory measurement method, but also significantly improves time efficiency.

Keywords

RF pulse design; eddy current; parallel transmission; high field

1. Introduction

Multidimensional spatially selective RF pulses [1, 2] have been proposed in a wide range of MRI applications, including localized excitation [3], spatially selective saturation [4], B_1 inhomogeneity reduction [5, 6] and B_0 inhomogeneity mitigation [7]. However, one major drawback of multidimensional RF pulses is their long pulse duration. For such long pulses, the effects of relaxation and off-resonance can significantly limit their practical use, especially at high field due to the presence of shorter transverse relaxation time and larger B_0 inhomogeneity. Recently, parallel transmission (PTX) [8-10] has been demonstrated as one effective means to overcome these limitations, as it allows the shortening of the RF pulse duration without sacrificing the quality of the spatial excitation [11, 12].

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The performance of parallel transmission can, however, be severely impeded by hardware and/or system imperfections. One of such imperfections is the effect of the eddy current field. Previous work with conventional single channel transmission systems has demonstrated that eddy currents can be produced by the gradient system and various methods, such as shielded gradients [13], pre-emphasis [14] and k-space trajectory measurement [15], have been proposed to eliminate or minimize their effects during RF excitation and imaging. One of the less discussed sources of eddy currents is from the RF shield when fast switching gradients are used. The RF shield, such as those used in a transverse electromagnetic (TEM) coil, can significantly improve the transmitter efficiency and reduce the specific absorption rate (SAR), which is particularly relevant for ultra-high field applications [16]. RF shield design has been proposed to reduce eddy currents [17]. Nevertheless, the effect of the remnant eddy current field can still introduce significant errors in parallel excitation that relies on the actual k-space trajectory, such as spiral or echo planar imaging (EPI). Wu *et al* [18] recently proposed and demonstrated the use of a k-space trajectory measurement to reduce the eddy currents effects for parallel excitation applications. However, since the k-space trajectory has to be re-measured whenever the nominal gradient waveform is modified, this approach is somewhat limited for use in routine PTX applications.

In this work, we demonstrate a model-based eddy current compensation method that relies on eddy current characterization [19-21] for mitigating their effects to improve the performance of PTX RF pulse design as well as accounting for potential system delays. With this approach, eddy current characterization needs to be performed only one time. Once the eddy current field is characterized, the eddy current distortions for any given gradient waveforms can be subsequently predicted using the established model and RF pulse design to provide the required compensation. In what follows we describe the methodology used for the characterization of the eddy current field and RF pulse design approach for compensating the eddy current effects. The performance of the proposed algorithm is then demonstrated on phantom studies using a Siemens 7 Tesla whole body scanner equipped with an eight-channel PTX system and also compared between unshielded and shielded RF coils.

2. Theory

2.1. Eddy current model and measurement

The eddy current field can be characterized as the superposition of multiple exponential terms, which represent the Taylor expansion of eddy current field. Higher-order terms are usually not considered due to their insignificant effects compared to linear terms and also difficult to measure and compensate [21,22]. The effect of eddy current field is spatially dependent and can severely distort the k-space trajectory. In addition to the effect from eddy currents, system delay exists between gradient waveforms and RF pulses. This delay can also hinder the performance of PTX excitation [18]. Therefore, RF pulses that are designed using an ideal k-space trajectory cannot always excite the desired spatial distribution.

According to previously described eddy current models [22], the eddy current field $G_E(t)$ induced by a nominal gradient waveform $G_D(t)$ can be described as the convolution of the eddy current impulse response function $H(t)$ and the negative time derivative $-dG_D(t)/dt$ of the nominal gradient waveform $G_D(t)$,

$$G_E(t) = - \frac{dG_D(t)}{dt} \otimes H(t) \quad (1)$$

where the eddy current impulse response function $H(t)$ can be modeled as the sum of multiple exponential terms [19, 20] that are characterized by amplitude constant α_n and time constant τ_n as follows,

$$H(t) = u(t) \sum_{n=0}^{N-1} \alpha_n e^{-t/\tau_n} \quad (2)$$

where $u(t)$ the unit step function. N is the number of multiple exponential terms.

The parameters of the model above can be determined using the measurement method of Atkinson *et al* [21] in conjunction with a constrained minimization algorithm. In short, data are acquired by applying calibration gradient waveforms multiple times with different gradient amplitudes and slice locations. In this study, we used one parameter of time delay and six parameters (i.e., three groups of amplitude constants and time constants) of multiple exponential terms (i.e. $N=3$) to express the impulse response function $H(t)$. We set the initial guesses of 0 for all the amplitude constants and 20 μ s, 50 μ s, and 100 μ s for three different time constants. The calibration gradient amplitude of 5mT/m with positive and negative directions were applied at seven different slice locations in each gradient direction. Therefore, the full set of system constants for each physical gradient are then obtained using an iterative nonlinear fitting method. With the system constants, we can estimate the impulse response function $H(t)$ according to Eq. (2). Then the eddy current gradient $G_E(t)$ can be predicted using Eq. (1). Finally, the actual gradient waveform can be obtained by summing up nominal gradient G_D and eddy current gradient G_E .

Parallel transmission RF pulses are designed in the small-tip-angle regime using the method proposed by Grissom *et al* [9] in which the linear small tip angle approximation [1] is extended for parallel transmission while controlling the RF power via regularization terms. Using this method, the excitation pattern ($\mathbf{m}(\mathbf{x})$) can be written to the desired RF filed as follows.

$$\mathbf{m}(\mathbf{x}) = i\gamma m_0 \sum_{r=1}^R S_r(\mathbf{x}) \int_0^T b_r(t) e^{ik(t) \cdot \mathbf{x}} e^{i\gamma \cdot \Delta B_0(\mathbf{x}) \cdot (t-T)} dt \quad (3)$$

where R is the number of coils; T is the pulse duration; B_0 is the spatial map of the static magnetic field inhomogeneity and S_r , $b_r(t)$ are the spatial sensitivity map (B_1^+ map) and RF pulse waveforms for coil r , respectively. The trajectory $k(t)$ is defined as the time-reversed integration of the gradient waveform. Equation (3) is then discretized in time and space and written in matrix form, $\mathbf{m}(\mathbf{x}) = \mathbf{A} \mathbf{b}(t)$. The RF pulses $\mathbf{b}(t)$ can then be obtained by solving the following minimization cost function,

$$\mathbf{b}(t) = \underset{\mathbf{b}}{\operatorname{argmin}} \left\{ \|\mathbf{A}\mathbf{b} - \mathbf{m}\|_{\mathbf{W}}^2 + \lambda \|\mathbf{b}\|^2 \right\} \quad (4)$$

where $\mathbf{m}$, as above, is the target magnetization; matrix $\mathbf{A}$ incorporates the excitation k-space trajectory, B_0 and B_1^+ maps; vector $\mathbf{b}(t)$ represents the complex RF pulse waveforms of all channels, $\mathbf{W}$ is a spatial error weighting mask that can be used to specify a region of interest (ROI) and the regularization parameter is used to balance contributions between RF power and excitation errors. This minimization problem is then solved using magnitude least square optimization method [23]. To improve the performance of the algorithm, correction terms are incorporated to compensate for the error associated with the finite gradient raster time [24]. Due to the mismatch between nominal and actual gradient waveforms, the corresponding nominal and actual trajectories will be obviously different. Therefore, the resulting uncorrected and corrected RF pulses will be significantly different.

Once the model parameters are determined, the performance of the proposed model-based method is tested by comparing the excitation patterns generated by RF pulses designed with the nominal gradient (G_D) and model-corrected actual gradient ($G_D + G_E$), respectively (regularization parameters () of 10 and flip angle of 10°) over a range of acceleration factors. In all cases, main magnetic field map correction is incorporated to exclude the effects of magnetic field inhomogeneities.

3. Materials and methods

3.1. B_1^+ and B_0 mapping

One of the most important ingredients for effective parallel transmission pulse design is the rapid and accurate mapping of the B_1^+ maps of the transmit coil array. The B_1^+ map (magnitude and phase) of each individual channel was acquired using a novel fast B_1^+ mapping method firstly introduced by Zhao *et al* [25]. This method can efficiently estimate B_1^+ maps for parallel transmission only using the images obtained from one small-tip-angle excitation. In this method, B_1^+ maps were obtained by sequentially transmitting through each RF channel while always receiving through all available receive channels. 2D GRE sequence with the following parameters: TR/TE = 200ms/2.51ms, flip angle = 10° , bandwidth per pixel = 1500 Hz/pixel, FOV = 220 mm and matrix size = 64 64 was used for this purpose (total scan time was 1 min 12 sec for mapping the B_1^+ of all channels). This B_1^+ mapping method produced accurate B_1^+ maps within very short imaging time, making the proposed PTX calibration practical for human experiments.

FLASH sequence with TE₁=3.7 ms and TE₂=4.7 ms, respectively, was used to estimate the B_0 map, which, as mentioned above, was purposely incorporated into the RF pulse design [23] in order to reduce the effect of off-resonance on the RF pulse design. This frequency offset compensation of field map proved particularly crucial in the context of ultra high field due to its large field inhomogeneity.

3.2. Eddy current effects on different coils

In order to illustrate eddy current effects between unshielded and shielded RF coils, the same experiments were performed using a decoupled birdcage coil (unshielded RF coil) and a TEM coil (shielded RF coil), respectively. Figure 1 shows the Siemens commercial birdcage coil and 4-channel transmit/receive TEM coil. Spiral and echo planar (EP) k-space trajectories were employed during the design of the spatially selective excitations. PTX acceleration was, therefore, achieved by undersampling the radial direction of the k-space trajectory in the spiral case and undersampling the blip direction of the k-space trajectory in the EP case. RF pulses were designed using nominal and model-corrected gradient waveforms, respectively.

3.3. Model-based method vs trajectory measurement method

We compared the excitation performance and time efficiency between the proposed model-based method and the direct k-space trajectory measurement method. The k-space trajectory was measured using the so-called peak finding technique [15,18] in which the k-space trajectory was extracted from the raw data measured after a brief phase encoding gradient was applied before playing out the desired k-space trajectory. The accuracy of this method relies on the number of encoding steps along the phase direction. More steps lead to higher accuracy but at the expense of increased acquisition time.

3.4 Numerical simulations

RF pulses were designed using spiral and EP excitation k-space trajectories, respectively. Spiral trajectory was designed with maximum gradient amplitude of 36 mT/m and

maximum slew rate of 150 mT/m/ms. PTX acceleration was achieved by undersampling the k-space trajectories in the radial direction. EP trajectory was designed with maximum gradient amplitude of 30 mT/m and maximum slew rate of 120 mT/m/ms. For the EP trajectory, acceleration was achieved by undersampling along the phase-encoding dimension. In both cases, the acceleration factor (R) was defined as the number of undersampling rate and approximately equal to the reduction rate of pulse duration. To demonstrate the effectiveness of the proposed method, acceleration factors of 1, 2, 3 and 4 for both spiral and EP trajectories were compared. When evaluating the performance of RF pulse design, the difference between the target and excited magnetizations was computed over the entire excited slice. Maximizing the spin excitation within the excited patterns and minimizing the spin excitation everywhere else were simultaneously implemented. We used the Normalized Root-mean-square error (NRMSE) defined as

$$\text{NRMSE} = \frac{1}{N_s} \sqrt{\sum_{i=1}^{N_s} |M_{xy}^D - M_{xy}^P|^2}$$

with ROI masked between the desired and excited patterns to evaluate the accuracy. All pulse designs and Bloch simulations were performed in MATLAB 2009a (Mathworks, Natick, MA).

3.5. Experiments

All experiments were performed on a Siemens 7T whole body scanner (Erlangen, Germany) equipped with an eight-channel PTX system and a gradient set with maximum amplitude of 40 mT/m and maximum slew rate of 170 mT/m/ms. The phantom used for the experimental data acquisition consisted of an 8 cm diameter sphere filled with 1.25 g/liter of nickel sulfate and 5 g/liter of sodium chloride.

For experimental data acquisition, spiral and EP designs with the same parameters as in numerical simulations for acceleration factors of 1, 2, 3 and 4 were compared. RF pulses designed with the nominal and model-corrected gradient waveforms were employed to excite a two-dimensional excitation pattern in the phantom. The desired pattern was a rectangle of $4.5 \times 9 \text{ cm}^2$ in the spiral case and a square of $4.5 \times 4.5 \text{ cm}^2$ in the EP case, respectively, on a 64×64 grid within a FOV of $20 \times 20 \text{ cm}^2$. 3D GRE sequence was used with the following parameters: 64×64 matrix, FOV = $20 \times 20 \text{ cm}^2$, TR = 30 ms, TE = 2.48 ms, 3.0 mm slice thickness and 300 Hz/pixel bandwidth.

4. Results

Figure 2 presents the excitation patterns produced by the unshielded birdcage coil with the spiral and EP trajectories, respectively. The results at the top row were obtained with the uncorrected RF pulses designed using the nominal gradient, while those at the bottom row were obtained with the corrected RF pulses designed using the model-corrected gradients. Clearly, the results are comparable due to the relatively insignificant effects of the eddy currents induced on this coil and also indicate that the eddy currents and delays of gradient system are negligible for our 7T PTX system. By contrast, the TEM coil generates stronger eddy currents due to the RF shield rendering significant eddy current induced errors. Because both gradient waveforms and k-space trajectory are distorted by the eddy current field (Fig. 3 a-b), pulses calculated on the actually generated trajectory, which is predicted by the model-based method, are significantly different from the pulses calculated on the nominal trajectory (Fig. 3c). Consequently, noticeable changes in the pulse shape lead to the correction of the excitation errors in both spiral (Fig. 5) and EP (Fig. 6) designs.

The comparison between the measured and model-corrected gradient waveforms for the spiral design is presented in Fig. 4a. The gradient waveforms generated by trajectory measurement method and model-based method are very similar, yet considerably different

from the nominal gradient waveform. Due to the distortion of the nominal gradient waveforms, RF pulses designed by the nominal gradient waveforms are inconsistent with the desired excitation pattern. However, both trajectory measurement method and model-based method produced good results. The excitation pattern obtained using the model-based method (Fig. 4c) seems to be slightly better than the one obtained using the trajectory measurement method (Fig. 4b).

Figure 5 shows the excitation patterns generated by the uncorrected pulses designed using nominal gradient and the corrected pulses designed using model-based method, respectively, with the spiral design for both simulations (Fig. 5a and 5b) and experiments (Fig. 5c and 5d). Notable tilt of the target pattern and outside artifacts were observed with the excitation of uncorrected RF pulses (Fig. 5a and 5c). It should be noticed that the distorted rotation primarily comes from the eddy current effects not from the delay between gradients and RF pulses, because it does not appear when we use the unshielded birdcage coil (Fig. 2). The performance of excitation was significantly improved when the uncorrected pulses (Fig. 5a and 5c) were replaced by the corrected pulses designed via model-based method (Fig. 5b and 5d). This behavior was consistent over the range of acceleration factors shown in this Fig. 5. The similar results were observed in the EP design (Fig. 6). The results obtained using the corrected pulses designed by the model-corrected method (Fig. 6b and 6d) provided improvements over those obtained using the uncorrected pulses designed by the nominal waveforms (Fig. 6a and 6c). Noticeably, there were considerable reductions of the ghosting artifacts along the phase-encoding direction due to the use of corrected pulses. Note that the excitation patterns at $R=1$ in Fig. 6c and 6d were contaminated by the long pulse duration ($>20\text{ms}$). This is also one application where parallel transmission can be applied to achieve improved excitation by shortening pulse duration.

In order to evaluate the excitation accuracy, we compared the normalized root-mean-square error (NRMSE) of uncorrected pulses and corrected pulses excited patterns over a range of acceleration factors for both spiral and EP designs (Fig. 7). In Fig. 7a, it was shown that smaller NRMSE could be achieved with the corrected RF pulses designed by model-based method over the entire range of acceleration factors. Moreover, when the effect of large B_0 field inhomogeneity was excluded, NRMSE could be further reduced for both excitations from uncorrected and corrected pulses. The graph in Fig. 7b demonstrated the similar behavior as in Fig. 7a. Note that the minimal NRMSE on the blue solid line of uncorrected pulses was not at $R=1$ but at $R=2$. However, when we excluded the effect of large field inhomogeneity, it presented that the minimal NRMSE was at $R=1$. The abnormal phenomenon was caused by the long pulse duration at the EP design ($>20\text{ms}$) as mentioned above, which suffered from the effects of both eddy current field and large field inhomogeneity.

5. Discussion and conclusion

To date, various methods have been proposed from hardware and software perspectives to reduce the undesired eddy currents induced by gradient switching. From the hardware perspective, Alecci *et al* [17] found that effective RF shielding and low eddy current sensitivity could be achieved by axial segmentation of a relative thick copper shield, etched on a kapton polyimide substrate. The research for reducing eddy currents with different shielding method is still ongoing in our center. However, our method is helpful not only for the reduction of the eddy currents on the RF shield, but also eddy currents from the gradient system and other delays due to hardware imperfections. While the eddy current effects from our Siemens gradient system are insignificant, other gradient system such as insert gradient system that has much stronger gradient amplitude and slew rate may gain benefits from the model-based procedure as reported here. From the software perspective, we have compared

the performance of trajectory measurement method with model-based method. Our proposed method achieves comparable or improved excitation quality. Trajectory measurement method does not lead themselves for practical applications due to the need of re-measuring the trajectory for any new trajectories in order to optimize the generation of a desired excitation pattern. While each trajectory measurement takes about 5-6 minutes, the acquisition time will be significantly increased when many trajectories are needed (e.g., when different accelerations are required) in one PTX session. The method demonstrated here circumvents this problem by using an established model to characterize the effect of the eddy currents on k-space trajectory. The model parameters need only be measured one time for a specific coil and the associated trajectory distortion can be subsequently predicted for any excitation gradient waveforms.

Good agreement is observed between numerical simulations and experimental results for both spiral and EP designs. Note, however, that these numerical simulations are strictly predicting the excitation pattern without the interference of hardware system, assuming homogeneous proton density and ignoring T_1 and T_2 relaxations. Moreover, large field inhomogeneity exacerbates the difficulties in achieving accurate excitation patterns in parallel transmission at 7T (Fig. 7). The impacts appears to be severe at $R=1$ due to the long pulse duration but limited for the relatively short duration of the accelerated RF pulses (Fig. 6). Though the B_0 map was incorporated in Eq. (3) for our studies, we have found that it is not adequate to employ the B_0 correction method in Eq. (3) for larger B_0 variation at 7T. To the best of our knowledge, for a given coil and gradient waveforms, the maximum of B_0 compensation will be determined for parallel transmission technique. Therefore, the compensation of B_0 variation tends to be deficient with the use of the reported B_0 correction method in Eq. (3) in the case of large B_0 inhomogeneity and/or long RF pulse duration due to large phase variation accumulation across different spatial locations. This is an interesting and practical topic of parallel transmission that is worthy of further investigations.

Consistent with others' works [18, 21, 26], the proposed method does not consider the effects of cross-terms related to the coupling of the gradient and/or RF coils, which leads to some remnant excitation errors. However, these could be accounted for through extensions of the methodology presented here.

In conclusion, we have successfully demonstrated that the model-based eddy current correction method using parallel transmission RF pulse design is an effective and efficient means to compensate for the excitation errors resulting from eddy currents on the RF shielding at 7T.

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References

- [1]. Pauly J, Nishimura D, Macovski A. A k-space analysis of small-tip angle excitation. *J. Magn. Reson.* 1989; 81:43–56.
- [2]. Pauly J, Nishimura D, Macovski A. A linear class of large-tip-angle selective excitation pulses. *J. Magn. Reson.* 1989; 82:571–587.
- [3]. Hardy CJ, Cline HE. Spatial localization in two dimensions using NMR designer pulses. *J. Magn. Reson.* 1989; 82:647–654.
- [4]. Pauly J, Hu B, Wang S, Nishimura D, Macovski A. A three-dimensional spin-echo or inversion pulse. *Magn. Reson. Med.* 1993; 29:2–6. [PubMed: 8419739]

- [5]. Saekho S, Boada FE, Noll DC, Stenger VA. Small tip angle three-dimensional tailored radio frequency slab-select pulse for reduced B_1 inhomogeneity at 3T. *Magn. Reson. Med.* 2005; 53:479–484. [PubMed: 15678525]
- [6]. Saekho S, Yip CY, Noll DC, Boada FE, Stenger VA. Fast-kz three-dimensional tailored radio frequency pulse for reduced B_1 inhomogeneity. *Magn. Reson. Med.* 2006; 55:719–724. [PubMed: 16526012]
- [7]. Stenger VA, Boada FE, Noll DC. Three-dimensional tailored RF pulses for the reduction of susceptibility artifacts in T_2^* -weighted functional MRI. *Magn. Reson. Med.* 2000; 55:719–724. [PubMed: 16526012]
- [8]. Katscher U, Bornert P, Leussler C, van den Brink JS. Transmit SENSE. *Magn. Reson. Med.* 2003; 49:144–150. [PubMed: 12509830]
- [9]. 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. *Magn. Reson. Med.* 2006; 56:620–629. [PubMed: 16894579]
- [10]. Zheng H, Zhao T, Qian Y, Ibrahim TS, Boada FE. Improved large tip angle parallel transmission pulse design through a perturbation analysis of the Bloch Equation. *Magn. Reson. Med.* 2011; 66:687–696. [PubMed: 21520270]
- [11]. 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; 56:1163–1171. [PubMed: 17036289]
- [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 B_1^+ inhomogeneity mitigation at 7 Tesla using parallel RF excitation with 16-element coil. *Magn. Reson. Med.* 2008; 60:1422–1432. [PubMed: 19025908]
- [13]. Chapman B, Mansfield P. Double active magnetic screening of coils in NMR. *J. Phys.* 1986; E19:L129.
- [14]. Wysong RE, Radio DP, Lowe IJ. A novel eddy current compensation scheme for pulsed gradient systems. *Magn. Reson. Med.* 1994; 31:572–575. [PubMed: 8015414]
- [15]. Takahashi A, Peters T. Compensation of multi-dimensional selective excitation pulses using measured k-space trajectories. *Magn. Reson. Med.* 1995; 34:446–456. [PubMed: 7500885]
- [16]. Snyder, C.; DelaBarre, L.; Van de Moortele, P.; Snyder, A.; Akgun, C.; Tian, J.; Metzger, G.; Ugurbil, K.; Vaughan, J. Stripline/TEM Transceiver Array for 7T Body Imaging; Proceedings of the 15th Scientific Meeting, International Society for Magnetic Resonance in Medicine; Berlin, Germany. 2007; p. 164
- [17]. Alecci M, Jezzard P. Characterization and reduction of gradient-induced eddy current in the RF shield of a TEM resonator. *Magn. Reson. Med.* 2002; 48:404–407. [PubMed: 12210952]
- [18]. Wu X, Vaughan J, Ugurbil K, Van de Moortele P. Parallel excitation in the human brain at 9.4T counteracting k-space errors with RF pulse design. *Magn. Reson. Med.* 2010; 63:524–529. [PubMed: 20017161]
- [19]. Jehenson P, Westphal M, Schuff N. Analytical method for the compensation of eddy current effects induced by pulses magnetic field gradients in NMR system. *J. Magn. Reson.* 1990; 90:264–278.
- [20]. Vanvaals JJ, Bergman AH. Optimization of eddy current compensation. *J. Magn. Reson.* 1990; 90:52–70.
- [21]. Atkinson IC, Lu A, Thulborn KR. Characterization and correction of system delays and eddy currents for MR imaging with ultrashort echo-time and time-varying gradients. *Magn. Reson. Med.* 2009; 62:532–537. [PubMed: 19353662]
- [22]. Bernstein, MA.; King, KF.; Zhou, XJ. Handbook of MRI pulse sequences. Academic Press; Boston: 2004.
- [23]. Setsompop K, Wald LL, Alagappan V, Gagoski B, Adalsteinsson E. Magnitude least squares optimization for parallel radio frequency excitation design demonstrated at 7 tesla with eight channels. *Magn. Reson. Med.* 2008; 59:908–915. [PubMed: 18383281]
- [24]. Zhao, T.; Zheng, H.; Hue, YK.; Ibrahim, TS.; Qian, Y.; Boada, FE. Practical consideration for the design of parallel transmission pulses at ultra high field; Proceedings of the 19th Scientific

Meeting, International Society for Magnetic Resonance in Medicine; Montreal, Canada. 2011; p. 4439

- [25]. Zhao, T.; Zheng, H.; DeFranco, A.; Ibrahim, TS.; Qian, Y.; Boada, FE. A fast B_1^+ mapping method for transmit/receive coils for parallel transmit (PTX) applications; Proceedings of the 19th Scientific Meeting, International Society for Magnetic Resonance in Medicine; Montreal, Canada. 2011; p. 2925
- [26]. Schneider, J.; Hass, M.; Hennig, J.; Junge, S.; Ruhm, W.; Ullmann, P. Coping with off-resonance effects and gradient imperfections in parallel transmission experiments; Proceedings of the 17th Scientific Meeting, International Society for Magnetic Resonance in Medicine; Honolulu, USA. 2011; p. 172

Highlights

- ▶ Model eddy current field induced on RF coils.
- ▶ Obtain system constants of eddy current via iterative nonlinear fitting method.
- ▶ Incorporate eddy current correction into parallel transmit RF pulse design at 7T.
- ▶ Improve efficiency and precision compared to trajectory measurement method.

Fig. 1. Different coils to illustrate different eddy current effects between unshielded and shielded RF coils. (a) Siemens commercial birdcage coil (unshielded RF coil) and (b) TEM coil (shielded RF coil).

Fig. 2. Excitation patterns produced by the Siemens birdcage coil using spiral and EP trajectories with an acceleration factor of 2. (a) Excitation obtained with RF pulses designed using the nominal gradient waveforms. (b) Excitation obtained with RF pulses designed using the model-corrected gradient waveforms. Note that no significant differences are observed, which implies that the eddy currents produced on the coil are negligible.

Fig. 3.

Gradient waveforms, k-space trajectory and resulting RF pulses for an acceleration factor of 2 with the use of the TEM coil. (a) Nominal vs actual (model-based correction) gradient waveforms. (b) Corresponding k-space trajectories. (c) Sum of amplitudes of all RF pulses obtained with the nominal (uncorrected) and actual (corrected) trajectories. Note that the deviation between the nominal and actual gradient waveforms leads to significant difference between the uncorrected pulses and corrected pulses.

Fig. 4.

Comparison of gradient waveforms and excitation patterns between model-based method and trajectory measurement method. (a) Gradient waveforms obtained from the nominal trajectory, the model-based method and the trajectory measurement method. The zoom-in panel clearly presents that the gradient waveforms derived from the model-based method and the trajectory measurement methods are very close. By contrast, these gradient waveforms are both significantly different from the nominal gradient waveforms. (b) Excitation obtained using the trajectory measurement method. (c) Excitation obtained using the model-based method.

Fig. 5.

Comparison of RF performance for the spiral trajectory design over a range of acceleration factors for both simulations (a-b) and experiments (c-d). (a) Simulated excitation patterns obtained with the RF pulses designed using the nominal gradient waveforms. (b) Simulated excitation patterns obtained with the RF pulses designed using the model-corrected gradient waveforms. (c) Experimental excitation patterns obtained with the RF pulses designed using the nominal gradient waveforms. (d) Experimental excitation patterns obtained with the RF pulses designed using the model-corrected gradient waveforms. Significant improvements are observed when the model-corrected RF pulses are used for both simulations and experiments.

Fig. 6.

Comparison of the RF performance for the EP trajectory over a range of acceleration factors for both simulations (a-b) and experiments (c-d). (a) Simulated excitation patterns obtained with the RF pulses designed using the nominal gradient waveforms. (b) Simulated excitation patterns obtained with the RF pulses designed using the model-corrected gradient waveforms. (c) Experimental excitation patterns obtained with the RF pulses designed using the nominal gradient waveforms. (d) Experimental excitation patterns obtained with the RF pulses designed using the model-corrected gradient waveforms. The result documents the same findings as in Fig. 5. Noticeably, significant reductions in ghosting artifacts along the phase encoding direction are obtained through the use of the proposed model-based correction method.

Fig. 7.

Comparison of excitation accuracy for (a) spiral design and (b) EP design over a range of acceleration factors in Bloch simulations. Note that the corrected RF pulses designed by model-based method increase excitation accuracy over the entire range of acceleration factors for both spiral and EP designs. The excitation accuracy is further increased when we exclude the effect of large B_0 field inhomogeneity. However, one interesting finding is that the minimal NRMSE on the solid blue line of uncorrected pulses in Fig. 7b is not at $R=1$ but at $R=2$. This is most probably due to the long pulse duration at the EP design ($>20\text{ms}$), which suffers from the severe effects of eddy current field and large field inhomogeneity.