Published in final edited form as:

IEEE Trans Med Imaging. 2014 August ; 33(8): 1641–1647. doi:10.1109/TMI.2014.2320987.

Iterative Method for Predistortion of MRI Gradient Waveforms

Kevin D. Harkins

Vanderbilt University Institute of Imaging Science, Vanderbilt University, Nashville, TN 37232 USA

Mark D. Does [Member, IEEE]

Departments of Biomedical Engineering, Electrical Engineering, and Radiology and Radiological Sciences, and the Vanderbilt University Institute of Imaging Science, Vanderbilt University, Nashville, TN 37232 USA

William A. Grissom* [Member, IEEE]

Departments of Biomedical Engineering, Radiology and Radiological Sciences, Electrical Engineering, and the Vanderbilt University Institute of Imaging Science, Vanderbilt University, Nashville, TN 37232 USA

Abstract

The purpose of this work is to correct for transient gradient waveform errors in magnetic resonance imaging (MRI), whether from eddy currents, group delay, or gradient amplifier nonlinearities, which are known to affect image quality. An iterative method is proposed to minimize error between desired and measured gradient waveforms, whose success does not depend on accurate knowledge of the gradient system impulse response. The method was applied to half-pulse excitation for 2-D ultra-short echo time (UTE) imaging on a small animal MRI system and to spiral 2-D excitation on a human 7T MRI system. Predistorted gradient waveforms reduced temporal signal variation caused by excitation gradient trajectory errors in 2-D UTE, and improved the quality of excitation patterns produced by spiral excitation pulses. Iterative gradient predistortion is useful for minimizing transient gradient errors without requiring accurate characterization of the gradient system impulse response.

Keywords

Artefacts; eddy currents; excitation; gradients; magnetic resonance imaging (MRI); predistortion

I. INTRODUCTION

GRADIENT trajectory errors are known to significantly degrade image quality in magnetic resonance imaging (MRI), and can be caused by eddy currents, group delay, and transient errors in gradient amplifier performance. The majority of MRI sequences use constant gradient pulses for slice selection and readout of Fourier-encoded image signals, which

minimizes the impact of gradient trajectory errors on image quality. Also, MRI gradient systems generally include preemphasis modules, which are based on a discrete decaying exponential model for the eddy currents, and pre-distort gradient amplifier waveforms to minimize the effects of eddy currents on image quality. However, various signal acquisition [1] and radio-frequency (RF) excitation [2]–[4] strategies that use time-varying gradient waveforms remain sensitive to inaccurate gradient waveform production, particularly when RF is played out concurrently with time-varying gradient waveforms, or when the center of k-space is sampled concurrently or shortly after gradient amplitude changes.

Several methods exist to measure actual gradient waveforms, and reconstructing images using those measured waveforms significantly improves image quality [5]–[8]. Similarly, during RF excitation, the excitation pattern or profile can be improved when the RF pulse is designed using the measured gradient waveforms [9], [10]. However, such retrospective corrections are not always acceptable. For example, in multidimensional parallel transmit MRI, for a given pulse duration the best overall excitation accuracy is achieved by optimizing RF and gradient waveforms jointly [11], [12], which precludes post-correction of the RF to match the actual realized gradient waveforms. In fact, the degraded conditioning of the parallel transmit pulse design problem caused by gradient waveform errors can lead to reduced selectivity and/or increased SAR [13]–[15]. In non-Cartesian imaging, eddy currents can lead to undersampled regions of k-space and subsequent aliasing artifacts that are not corrected by reconstruction on the measured trajectory. Also, half-pulse excitation for 2-D ultrashort echo-time (UTE) MRI involves two acquisitions, which are digitally combined to cancel out-of-slice signal. Any residual gradient amplitude after the RF excitation can alter the slice profile in a time-dependent manner, causing out-of-slice signal to contaminate the image [16].

Some research has aimed to characterize the gradient system, in order to estimate true gradient waveforms or improve gradient performance. Most approaches assume a linear time-invariant (LTI) approximation to the gradient system response, and some use test waveforms to explicitly characterize the LTI impulse response [17], [18]. The resulting gradient system models have mostly been used to estimate the actual gradient waveforms for proper reconstruction of image data, but they have also been used to predistort slice-select gradient waveforms to improve short-time constant eddy currents which cause 2-D UTE slice profile errors [19].

A method was recently introduced that iteratively predistorts gradient waveforms to minimize the error between measured and ideal waveforms [20]. The present work improves this iterative technique using a regularized inversion of an estimate of the impulse response. The resulting method is less sensitive to the accuracy of the estimated impulse response compared to LTI based methods, and is based on an established iterative control method known to converge even for nonlinear systems [21]. The method is demonstrated for suppressing out-of-slice signal contamination in 2-D UTE imaging, and for improving the accuracy of a spiral RF excitation.

II. THEORY

Iterative learning control (ILC) is a branch of control theory that attempts to optimize a set of input parameters such that the output of a system matches a desired action, and is applicable when the system response is not precisely known or may even be nonlinear [21]. In MRI, gradient waveforms represent a repeatable system, where a length- N_t applied input waveform, $\mathbf{g}_{app} = \{G_{app}(t_i)\}_{i=1}^{N_t}$, can be adjusted and designed such that the corresponding measured waveform, $\tilde{\mathbf{g}}$, converges to an ideal waveform, $\mathbf{g}_{nom}$. An update rule is defined for the applied gradient waveform as

$$\mathbf{g}_{app}^{j+1} = \mathbf{g}_{app}^j + \mathbf{K} \Delta_g^j \quad (1)$$

where j is the iteration number, $\mathbf{K}$ is an update matrix, and Δ_g^j is the measured error at iteration j

$$\Delta_g^j = \mathbf{g}_{nom} - \tilde{\mathbf{g}}^j. \quad (2)$$

For an LTI system and perfect measurements, the problem can be solved in one iteration using $\mathbf{K} = \mathbf{A}^{-1}$, where $\mathbf{A}$ is a Toeplitz matrix whose rows are shifted copies of the impulse response of the gradient system. This would be equivalent to previously-published correction methods that assume an LTI system. However, in practice, the impulse response can be difficult to accurately characterize, inversion of $\mathbf{A}$ is generally unstable, and the gradient amplifier performance is transiently nonlinear. When considering such a system, constrained iterative solutions can result in a stable optimization. For this work, $\mathbf{K}$ is chosen to be the regularized pseudoinverse of an estimated matrix $\tilde{\mathbf{A}}$

$$\mathbf{K} = \kappa (\tilde{\mathbf{A}}^T \tilde{\mathbf{A}} + \lambda \mathbf{I})^{-1} \tilde{\mathbf{A}} \quad (3)$$

where the rows of $\tilde{\mathbf{A}}$ are shifted copies of the measured impulse response of the gradient system. The regularization parameter λ balances conditioning of the inversion with increasing the number of iterations needed for convergence and possibly biasing the optimized solution. The parameter κ is a step size, and trades off stability and insensitivity to noise with the number of iterations required for convergence. The normalized root mean square error (NRMSE), calculated as

$$NRMSE^j = \frac{\|\Delta_g^j\|_2}{\|\mathbf{g}_{nom}\|_2} \quad (4)$$

is used to measure the gradient error at each iteration j , and the procedure is considered converged when the NRMSE increases compared to the previous iteration. We note that this method's ability to significantly improve excitation accuracy is predicated on the assumption that field distortions are dominated by zeroth- and first-order terms, and can thereby be alleviated using predistortion.

The method described above is not guaranteed to produce gradient waveforms that obey gradient amplitude and slew rate limits. If the target waveforms $\mathbf{g}_{nom}$ come close to those

limits, it may be desirable to enforce them at each iteration. One way to do this is to solve the following quadratic program:

$$\begin{aligned} & \text{minimize} && \|g_{app}^{j+1} - \hat{g}\|^2 \\ & \text{subject to} && -a_{max} \leq \hat{g} \leq a_{max} \quad (5) \\ & && D\hat{g} \leq s_{max}\Delta_t \end{aligned}$$

where g_{app}^{j+1} is the nominal applied waveform for iteration $j + 1$ [(1)], $\hat{g}$ is the amplitude- and slew rate-limited waveform to be solved for, a_{max} is the amplitude limit, D is a first-order finite- differencing matrix, s_{max} is the slew rate limit, and δ_t is the dwell time with which the gradient waveform is sampled. Performing this calculation for each gradient channel will produce new applied gradient waveforms that, within the set of all waveforms meeting amplitude and slew rate constraints, are closest to the nominal applied waveforms g_{app}^{j+1} in a least-squares sense.

III. METHODS

The proposed predistortion method and all other calculations were implemented in MATLAB R2013a (Mathworks, Natick, MA, USA). MATLAB's `convmtx()` function was used to build the A matrices, and the backslash (“\”) operator was used to calculate K [(3)].

A. Half-Pulse UTE Imaging

A bipolar slice selection gradient pulse was designed to minimize the RF excitation time in a 2-D UTE radial imaging sequence, based upon a 40 Gauss/cm limit on gradient amplitude, a 100 Gauss/cm/ms limit on slew rate and a 2000Gauss/cm/ms² limit on acceleration. A 20°, 0.4 ms Gaussian RF half pulse was used for excitation. It was interpolated to the measured gradient waveforms [3], and compensated for measured time-varying B_0 offsets by applying the phase shifts induced by those offsets to the pulse. Residual B_0 eddy currents during the readout created by the excitation gradient waveforms resulted in phase offsets of less than 0.1 radians, and were considered to be negligible. A multi-slice variant of a previously described method [8] was used to measure the gradient and B_0 waveforms. Gradient cross-component terms, such as the x gradient fields induced by z gradient pulses, were not measured and were assumed to be negligible. RF and gradient waveforms were sampled with a 4 μ s dwell time for scanner implementation.

UTE images were collected on a 7T, 16 cm horizontal bore magnet with Varian/Agilent DirectDrive console (Santa Clara, CA, USA) using a 38-mm Litz quadrature coil (Doty Scientific, Columbia, SC, USA). Signals from two acquisitions, distinguished by the polarity of the slice select gradient waveforms, were collected and digitally combined to cancel the out-of-slice signal induced by each half pulse. For each pass, 256 dummy scans were used to ensure magnetic steady state, and 2-D images were acquired using 128 readout points along 256 radial views with a 250 kHz receiver bandwidth and two averaged excitations per pass. The raw signal was density compensated [22]–[24] and reconstructed to a Cartesian grid with 5×5 cm field-of-view and 128×128 matrix size using a nonuniform fast Fourier transform [25] and a measured k-space trajectory [8]. Half-pulse images were collected before and after slice-select gradient waveform optimization. The impulse response of the z

gradient channel was measured using the same method as was used for the slice-select waveforms, and a slew rate-limited, 0.4 Gauss/cm impulses, each comprising a single 4 μ s sample. The measured impulse response was then directly used in the $\mathbf{A}$ matrix, without further processing. The scanner's slew rate limit was 200 Gauss/cm/ms, so amplitude and slew rate limiting were not used in the predistortion method. To illustrate the effect of eddy currents after half pulse excitation on the slice profile, images of a CuSO_4 solution ($T_1 \approx T_2 \approx 300$ ms) were collected at 20 echo times logarithmically spaced between 0.1 and 2 ms. The acquisition time for a single image was 30 s.

B. Spiral Excitation

A 12.7 ms spiral-in excitation pulse was designed to excite the desired pattern shown in Fig. 2(a). The trajectory was designed using the method in [26], with a slew rate limit of 12 Gauss/cm/ms, and was slew rate-constrained for its entire duration. The trajectory reached a maximum gradient amplitude of 2.16 Gauss/cm, had 16 turns, and encoded a 10-cm FOV and 3-mm resolution. The target excitation pattern was sampled on a 128×128 grid, with a target flip angle of 90° . The corresponding RF pulse was designed using a conjugate gradient-based algorithm [27], with nonuniform fast Fourier transforms to evaluate system matrix multiplications [25]. The design used 100 iterations and a roughness penalty to encourage smooth real and imaginary RF waveforms [27]. The RF and gradient waveforms were sampled with a 6.4 μ s dwell time for scanner implementation.

Predistortion experiments were performed on a 7T Philips Achieva whole-body scanner (Philips Healthcare, Cleveland, OH, USA). The method of [8] was used for all waveform measurements. Gradient cross-component terms were not measured and were assumed to be negligible. First, the impulse responses of the x and y gradient channels were measured using 0.3 ms triangular gradient pulses with 12 Gauss/cm/ms slew rate on their ramps, and peak amplitudes of 1.67 Gauss/cm. The impulse responses were determined from the measured gradient waveforms by deconvolving the input triangular pulses from them via regularized division in the Fourier domain. Predistortion was run using the impulse responses and step sizes of $\kappa = 0.5$ and 1, and a third time assuming the impulse response was a delayed impulse with $\kappa = 0.5$, where the delay was estimated using the algorithm in [28]. $\lambda = 0.01$ was used in all three runs. The scanner's amplitude and slew rate limits were 4 Gauss/cm and 18 Gauss/cm/ms. The nominal predistorted waveforms came close to or violated the slew rate limit near the center of k -space, so they were constrained by solving the quadratic program in (5). That optimization was implemented using MATLAB's `quadprog()` function, and the matrix $\mathbf{D}$ was implemented as a sparse matrix. Excitation patterns were imaged in a ball phantom using a 2DFT spin echo sequence in which the spiral pulses were used for excitation, with a slice-selective refocusing pulse for selection in the z -dimension. The sequence had parameters $TE = 5.35$ ms (measured from the end of the spiral waveform), $TR = 500$ ms, four averages, resolution 0.9 mm. A two-sample delay between the RF and gradient channels was measured and corrected. An additional image was collected with a conventional 90° slice-selective excitation pulse to divide out the effects of transmit and receive RF field inhomogeneity.

IV. RESULTS

A. Half-Pulse UTE Imaging

Fig. 1 shows the half-pulse UTE imaging results. Fig. 1(a) plots the nominal, uncorrected, and group delay-corrected-only gradient waveforms, which were measured using the same method that was used for predistortion. The uncorrected waveform is visibly delayed. When that group delay is compensated, other errors become more visible: the waveform does not reach the nominal maximum and minimum, and it remains high after the start of the TE period at 0 ms. The predistorted waveform would be indistinguishable from the nominal waveform in this view. Fig. 1(b) shows a zoomed-in view at the end of the pulse and start of the TE period, and plots the nominal, group delay-corrected-only, and predistorted waveforms. Iterative predistortion effectively suppresses the residual gradient after the start of the TE period. Fig. 1(c) plots gradient waveform error for uncorrected, group delay-corrected-only, and predistorted cases. Fig. 1(d) shows the convergence of the method in three different scenarios by plotting the NRMSE. An LTI-based predistortion approach that was previously proposed to improve half-pulse excitations [19] (realized here using $\kappa = 1$ and $\lambda = 0$) improved the gradient error by an order of magnitude, but was sensitive to noise, and the NRMSE diverged after a single iteration. Regularized inversion ($\lambda = 1$ curves) was more robust to noise, and upon convergence the NRMSE of the gradient error was more than an order of magnitude improved from the unregularized LTI predistortion. Iterating with a full step ($\kappa = 1$) versus a half step ($\kappa = 0.5$) made almost no difference in the converged solution, although the half step took longer to converge. The predistorted results in Fig. 1(b), (c), and (f) correspond to the half step solution.

In 2-D half-pulse UTE imaging the residual eddy currents in the uncorrected case affected the out-of-slice signal cancellation, resulting in a substantial variation in the signal amplitude as a function of echo time, as shown in Fig. 1(f). In contrast, images excited with predistorted gradient waveforms for half-pulse slice selection showed almost no temporal variability, indicating negligible residual gradients after the end of the prescribed gradient waveform.

B. Spiral Excitation

Fig. 2 shows the spiral excitation results. The original RF pulse and gradient waveforms excited a pattern with substantial blurring in three areas. The rms error between simulated and target excitation patterns was 23.9° when that RF pulse was simulated using the measured trajectory without predistortion. Redesigning the RF pulse on the measured, uncorrected gradient waveforms largely corrected the blurring in the measured pattern, and in simulation reduced the rms error to 11.3° . Iterative predistortion corrected the errors without requiring RF redesign, and in simulation reduced the rms error to 11.2° . These errors and the visual appearance of the measured patterns in Fig. 2(a) suggest that iteratively predistorting the waveforms and redesigning them on the measured trajectory resulted in equally-accurate excitations in this design scenario.

Fig. 2(b) shows gradient waveform and excitation k-space trajectory errors with and without predistortion, which were measured using the same method that was used for predistortion.

The gradient errors without predistortion were of relatively constant amplitude in the outer parts of excitation k -space, and could have been largely corrected with an input waveform amplitude adjustment. Near the center of k -space, though, the errors grew in amplitude due to an increasing delay between the nominal and realized waveforms. With predistortion, the errors on both x and y channels appear to have been reduced to the measurement noise level.

Fig. 2(c) shows NRMSE versus iteration for three predistortion cases. With $\kappa = 1$, predistortion with the measured impulse response reached an error of 0.94% after one iteration, but diverged in the second iteration. In comparison, with $\kappa = 0.5$, predistortion converged to a lower error of 0.5% after eight iterations, suggesting that the error was not entirely explained by the measured LTI gradient model. These two cases are compared with a previously-described predistortion approach for improving multidimensional excitations that does not use a measured impulse response [20], which converged more slowly to a higher error of 0.7% after nine iterations; with the impulse response this error was reached after four iterations. Fig. 2(d) plots the nominal and measured excitation k -space trajectories without and with predistortion, where the measured trajectories were obtained using the same method that was used for predistortion. The nominal and predistorted trajectories are indistinguishable.

V. Discussion

The ability to mitigate the effects of gradient trajectory errors is increasingly important as the speed of MR imaging is pushed to the limit using fast, non-Cartesian readout trajectories, and as multidimensional RF excitation techniques are developed to overcome RF transmit field inhomogeneity in ultra-high field MRI [29]. Especially for excitation schemes that are not compatible with retrospective design and correction of gradient waveforms, there is a need for prospective methods to predistort applied gradient waveforms such that the realized waveform matches a desired one. This work presented an iterative predistortion method to achieve that goal.

In the present work, predistortion was applied to gradient waveforms in two applications of RF excitation: half-pulse excitation for 2-D UTE imaging, and 2-D selective excitation using a spiral trajectory. In 2-D UTE imaging, gradient optimization minimized echo-time dependent signal variations by minimizing residual gradient error after a half-pulse excitation, which are known to cause out-of-slice signal contamination. The spiral excitation pattern was also significantly improved after reduction of gradient waveform error by predistortion; however, the best overall excitation accuracy would likely be attained by redesigning the RF pulse on the final predistorted gradient waveforms. Redesigning the pulse on the final predistorted waveforms would also enable compensation of higher-order field distortions [10], which were not addressed by the proposed method.

The proposed method was demonstrated to reduce gradient waveform errors without requiring accurate characterization of the system. For example, in half-pulse slice selection, regularization of the gradient waveform correction improved gradient error more after a single iteration compared to unregularized LTI-based predistortion, due to sensitivity of the unregularized update to noise. Further iteration of the regularized correction method

continued to improved gradient error, while the unregularized predistortion diverged after a single iteration. The spiral results further showed that the method converges faster with an accurate impulse response, but can still converge to substantially-improved gradient waveforms without it. The spiral gradient waveforms were optimized with two different estimates of the impulse response: one based upon an actual measurement, and another that was a simple shifted delta function, which could be considered the simplest impulse response to assume. While optimization with both of these estimates improved the gradient error, the correction that used the measured impulse response converged to a lower error. The method could also be capable of compensating the effects of amplifier nonlinearities, such as compression and hysteresis, that cannot be captured in an LTI model. Those effects may have played a role in the results presented here, and been one of the reasons why multiple predistortion iterations resulted in significantly lower error than a single LTI-based iteration, in both experiments.

Gradient cross-component terms (such as x gradient fields induced by y gradient pulses, and vice versa) were assumed to be negligible and were not measured in the present work. However, the method could be extended to account for them by reformulating the update equations as a joint update for all channels, wherein the three gradient channels' measured, nominal and applied waveforms and measured impulse responses are stacked. All waveform measurements would then need to be performed using a technique capable of resolving cross-components, which the slice-offset methods used here cannot measure accurately.

The best choices of the regularization parameter λ and the step size κ for predistortion will be application-dependent. The best λ to use will depend multiple factors including the shape of the measured impulse response and the relative noise level of the impulse response measurements, and it should be chosen to minimize the influence of noise without imparting bias to the solution. It is possible that λ could be chosen via automatic ridge parameter estimates like generalized cross validation [30]. Comparing the UTE imaging and 2-D spiral excitation results, it is apparent that the best step size κ is application-dependent: a full step size of 1 resulted in the best solution for the UTE imaging application on the small animal MRI system, whereas the partial step size produced the best solution for spiral excitation on the clinical MRI system.

The proposed method aims to iteratively adjust an input waveform to match a desired output waveform, and does not prescribe a generalized model for prospective gradient characterization and correction applicable to any desired waveform. As such, the method needs to be repeated whenever the desired waveform changes, whether that be a change in shape, duration, or amplitude. Depending on the transient nonlinearity of the gradient waveforms, even waveforms rotated to different axis or oblique angles will need to be optimized independently. However, it may be possible to generalize the method for system characterization using model order reduction techniques. Furthermore, the method could be accelerated using NMR field probes for trajectory measurement [31], [32], rather than the slower slice-offset methods used here. This would also provide higher SNR, which would lead to improved impulse response measurements and the ability to reach lower errors. Such accelerations and enhancements of the method would also facilitate its extension to improve non-Cartesian image reconstructions which, compared to multidimensional excitations,

usually use much longer gradient waveforms and multiple shots. Compared to retrospective trajectory corrections for non-Cartesian imaging, the method may provide improved conditioning of the reconstruction problem, which would translate to local improvements in SNR and artifact reductions, since the targeted k-space sampling density would be better achieved. The use of NMR field probes may also enable the measurement of higher-order and cross-component field terms.

In the present implementation of the algorithm the compute times were negligible (milliseconds) compared to waveform measurement times (several seconds). If the measurement times can be reduced then it may be desirable to also accelerate the computation. The dominant operation in each iteration is the multiplication by $\mathbf{K}$ in (1), which has complexity $O(N^2)$, where N is the number of samples in the gradient waveform. For long gradient waveforms, it may be faster to implement that multiplication using FFT's, since the matrix $\tilde{\mathbf{A}}$ is a convolution matrix. This would reduce the complexity to $O(N \log N)$.

VI. CONCLUSION

The proposed iterative predistortion method allows gradient waveforms to be optimized to match a desired waveform, and is useful in correcting gradient errors that can corrupt image quality, and in excitation applications where gradient and RF waveforms are jointly designed.

Acknowledgments

This work was supported by the National Institutes of Health (NIH) under Grant EB0014308, Grant DA019912-06, and Grant EB016695. *Asterisk indicates corresponding author.*

REFERENCES

1. Ahn C, Kim J, Cho Z. "High-speed spiral-scan echo planar NMR imaging," IEEE Trans. Med. Imag. Mar.5(1):2–7. 1986.
2. Pauly JM, Nishimura DG, Macovski A. A k-space analysis of small-tip-angle excitation. J. Magn. Reson. 1989; 81:43–56.
3. Hargreaves BA, Cunningham CH, Nishimura DG, Conolly SM. Variable-rate selective excitation for rapid MRI sequences. Magn. Reson. Med. Sep.2004 52(3):590–597. [PubMed: 15334579]
4. Schröder C, Börnert P, Aldefeld B. Spatial excitation using variable-density spiral trajectories. J. Magn. Reson. Imag. 2003; 18(1):136–141.
5. Alley MT, Glover GH, Pelc NJ. Gradient characterization using a Fourier-transform technique. Magn. Reson. Med. 1998; 39(4):581–587. [PubMed: 9543420]
6. 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]
7. Zhang Y, Hetherington HP, Stokely EM, Mason GF, Twieg DB. A novel k-space trajectory measurement technique. Magn. Reson. Med. 1998; 39:999–1004. [PubMed: 9621924]
8. Gurney, P.; Pauly, JM.; Nishimura, DG. A simple method for measuring B_0 eddy currents; Proc. 13th Sci. Meet., Int. Soc. Magn. Reson. Med.; Miami Beach, FL. 2005. p. 866
9. 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; Proc. 16th Sci. Meet., Int. Soc. Magn. Reson. Med.; Toronto, Canada. 2008. p. 1313

10. Schneider JT, Haas M, Ruhm W, Hennig J, Ullmann P. Robust spatially selective excitation using radiofrequency pulses adapted to the effective spatially encoding magnetic fields. *Magn. Reson. Med.* 2011; 65:409–421. [PubMed: 20872857]
11. Yip CY, Grissom W, Fessler JA, Noll DC. Joint design of trajectory and RF pulses for parallel excitation. *Magn. Reson. Med.* 2007; 58(3):598–604. [PubMed: 17763362]
12. Grissom WA, Khalighi MM, Sacolick LI, Rutt BK, Vogel MW. Small-tip-angle spokes pulse design using interleaved greedy and local optimization methods. *Magn. Reson. Med.* 2012; 68:1553–1562. [PubMed: 22392822]
13. Zhu Y. Parallel excitation with an array of transmit coils. *Magn. Reson. Med.* Apr.2004 51(4):775–784. [PubMed: 15065251]
14. Zhu, Y. RF power deposition and 'g-factor' in parallel transmit; Proc. 14th Sci. Meet., Int. Soc. Magn. Reson. Med.; Seattle, WA. 2006. p. 599
15. Lattanzi R, Sodickson DK, Grant AK, Zhu Y. Electrodynamics constraints on homogeneity and radiofrequency power deposition in multiple coil excitations. *Magn. Reson. Med.* 2009; 61(2): 315–334. [PubMed: 19165885]
16. Pauly JM. Selective excitation for ultrashort echo time imaging. *Encycl. Magn. Reson.* 2012; 1:1–7.
17. Vannesjo SJ, Haerberlin M, Kasper L, Pavan M, Wilm BJ, Barmet C, Pruessmann KP. Gradient system characterization by impulse response measurements with a dynamic field camera. *Magn. Reson. Med.* 2013; 69:583–593. [PubMed: 22499483]
18. Addy NO, Wu HH, Nishimura DG. Simple method for MR gradient system characterization and k-space trajectory estimation. *Magn. Reson. Med.* 2012; 68:120–129. [PubMed: 22189904]
19. Lu A, Daniel BL, Pauly JM, Pauly KB. Improved slice selection for R2* mapping during cryoablation with eddy current compensation. *J. Magn. Reson. Imag.* 2008; 28(1):190.
20. Grissom, WA.; Kerr, AB.; Stang, P.; Scott, GC.; Pauly, JM. GrIP: Gradient iterative predistortion for multidimensional and parallel excitation; Proc. 18th Sci. Meet., Int. Soc. Magn. Reson. Med.; Stockholm, Sweden. 2010. p. 4925
21. Ahn H-S, Chen Y, Moore KL. Iterative learning control: Brief survey and categorization. *IEEE Trans. Syst., Man Cybern. C.* Nov.2007 37(6):1099–1121.
22. Pipe JG, Menon P. Sampling density compensation in MRI: Rationale and an iterative numerical solution. *Magn. Reson. Med.* 1999; 41:179–186. [PubMed: 10025627]
23. Johnson KO, Pipe JG. Convolution kernel design and efficient algorithm for sampling density correction. *Magn. Reson. Med.* 2009; 61:439–447. [PubMed: 19165893]
24. Zwart NR, Johnson KO, Pipe JG. Efficient sample density estimation by combining gridding and an optimized kernel. *Magn. Reson. Med.* 2012; 67:701–710. [PubMed: 21688320]
25. Fessler JA, Sutton BP. Nonuniform fast Fourier transforms using min-max interpolation. *IEEE Trans. Signal Process.* Feb.2003 51(2):560–574.
26. Glover GH. Simple analytic spiral k-space algorithm. *Magn. Reson. Med.* Aug.1999 42(2):412–415. [PubMed: 10440968]
27. Yip CY, Fessler JA, Noll DC. Iterative RF pulse design for multidimensional, small-tip-angle selective excitation. *Magn. Reson. Med.* Oct.2005 54(4):908–917. [PubMed: 16155881]
28. Tomasi, C.; Shi, J. Good features to track; Proc. IEEE Conf. Comput. Vis. Pattern Recognit.; 1994. p. 593-600.
29. Grissom WA, Sacolick L, Vogel MW. Improving high-field MRI using parallel excitation. *Imag. Med.* 2010; 2(6):675–693.
30. Golub GH, Heath M, Wahba G. Generalized cross-validation as a method for choosing a good ridge parameter. *Technometrics.* 1979; 21:215–223.
31. De Zanche N, Barmet C, Nordmeyer-Massner JA, Pruessmann KP. NMR probes for measuring magnetic fields and field dynamics in MR systems. *Magn. Reson. Med.* 2008; 60(1):176–186. [PubMed: 18581363]
32. Schneider, JT.; Barmet, C.; Ruhm, W.; Pruessmann, KP.; Ullmann, P. Using dedicated field probes for trajectory measurements in parallel excitation experiments; Proc. 18th Sci. Meet. Int. Soc. Magn. Reson. Med.; Stockholm, Sweden. 2010. p. 4926

Fig. 1.

Half-pulse UTE imaging: (a) Gradient waveforms for nominal (grey line), measured uncorrected (black crosses) and measured group delay corrected-only (gray dots) cases. In this view, the measured predistorted gradient waveform would be indistinguishable from the nominal waveform. (b) Zoomed-in view of (a), at the end of the bipolar pulse and the start of the TE period. This view shows the nominal, group delay corrected-only, and predistorted waveforms. (c) Measured error between nominal and uncorrected, group delay corrected-only, and predistorted waveforms. (d) NRMSE versus iteration for LTI-based gradient correction (dashed line), regularized half-step (black line) and whole-step (gray line) iterative predistortion. (e) Representative reconstructed UTE image indicating in red the ROI over which signals plotted in (f) were averaged. (f) UTE image signal in the ROI without (gray) and with (black) predistortion.

Fig. 2.

Spiral-in excitation results. (a) Nominal, predicted, and measured excitation patterns. With no RF or gradient correction ("No Correction" case), there is substantial blurring of the excitation pattern due to trajectory errors. Redesigning the RF on the measured trajectory ("Redesigned RF" case) alleviates the errors, as does iterative predistortion ("Predistorted" case). (b) Measured errors of initial ("No predistortion") and predistorted gradient waveforms and excitation k-space trajectories. (c) Error versus iteration for three predistortion cases. Predistortion using the measured impulse response converged to the lowest error. (d) k-Space trajectory of measured gradients with and without predistortion. With predistortion, the nominal and measured trajectories appear coincidental.