Published in final edited form as:

Magn Reson Med. 2018 July ; 80(1): 101–111. doi:10.1002/mrm.27009.

Phase-Sensitive B_1 Mapping: Effects of Relaxation and RF Spoiling

Jie Wen, Alexander L. Sukstanskii, and Dmitriy A. Yablonskiy*

Department of Radiology, Washington University, St. Louis, Missouri, USA.

Abstract

Purpose: To develop a phase-based B_1 mapping technique accounting for the effects of imperfect RF spoiling and magnetization relaxation.

Theory and Methods: The technique is based on a multi-gradient-echo sequence with 2 successive orthogonal radiofrequency (RF) excitation pulses followed by the train of gradient echoes measurements. We have derived a theoretical expression relating the MR signal phase produced by the 2 successive RF pulses to the B_1 field and B_0 -related frequency shift. The expression takes into account effects of imperfections of RF spoiling and T_1 and T_2^* relaxations.

Results: Our computer simulations and experiments revealed that imperfections of RF spoiling cause significant errors in B_1 mapping if not accounted for. By accounting for these effects along with effects of magnetization relaxation and frequency shift, we demonstrated the high accuracy of our approach. The technique has been tested on spherical phantoms and a healthy volunteer.

Conclusion: In this paper, we have proposed, implemented, and demonstrated the accuracy of a new phase-based technique for fast and robust B_1 mapping based on the measured MR signal phase, frequency, and relaxation. Because imperfect RF spoiling effects are accounted for, this technique can be applied with short TRs and therefore substantially reduces the scan time.

Keywords

MRI; B_1 mapping; RF spoiling; flip angle mapping; frequency shifts

INTRODUCTION

An accurate measurement of B_1 radiofrequency (RF) field is important for many MRI applications, such as quantitative mapping of MR relaxation parameters, RF pulse design for parallel transmission (1–5), and mapping electric tissue properties (6–8). For example, mapping of a longitudinal relaxation rate constant R_1 often relies on MRI with variable flip angles (VFA) (9–16). However, the accuracy of VFA methods not only depends on the models (16–22) used to analyze data but also highly depends on accuracy of B_1 measurements.

*Correspondence to: Dmitriy A. Yablonskiy, Ph.D., Mallinckrodt Institute of Radiology, Washington University, 4525 Scott Ave., Room 3216, St. Louis, MO 63110. yablonskiyd@wustl.edu.

SUPPORTING INFORMATION

Additional Supporting Information may be found in the online version of this article.

Up to date, several methods have been proposed to measure B_1 , either based on the magnitude (23–33) or phase (34–38) of the MR signal. Initially, the doubleangle methods (23,24) were used for B_1 measurement. In these methods, long repetition times ($TR > 5T_1$) are necessary to eliminate the T_1 -dependence of the signal magnitude, therefore usually requiring long acquisition time. Substantial reduction of the dependence of B_1 evaluation on the system relaxation parameters has been achieved in the methods proposed by Yarnykh (30) and Nehrke et al. (32,33). As an alternative to magnitude-based methods, the MR signal phase-based methods have been proposed to measure B_1 field (34–37). To generate the MR signal phase dependence on flip angle, Morell (35) and later Chang (37) proposed to use a pair of successive orthogonal RF pulses, whereas Sacolick et al. (36) proposed to use the Bloch-Siegert phase shift effect (39).

In this paper, we further explore a phase-sensitive method for B_1 mapping that is based on a pair of successive orthogonal RF pulses encoding information on B_1 field strength. We demonstrate that the MR signal phase generated by the pair of successive orthogonal RF pulses depends not only on B_1 field strength but also on the MR signal relaxation properties (T_1 and T_2^*) and (most importantly) on imperfections of RF spoiling. These effects have not been considered previously. We have developed a theory accounting for contribution of the above mentioned effects (T_1 , T_2^* and RF spoiling) as well as the B_0 -induced frequency shift to the MR signal phase and proposed experimental method for B_1 mapping relying on a multi-gradient-echo sequence with 2 successive orthogonal RF pulses used for signal excitation. The use of the multi-gradient-echo sequence provides simultaneous measurements of the B_1 -encoded MR signal phase, signal transverse relaxation rate constant R_2^* and the B_0 -dependent MR signal frequency shift. The feasibility and accuracy of the method are examined using computer simulations and demonstrated on a phantom and on a healthy volunteer.

THEORY

Two consecutive RF pulses (corresponding to the nominal flip angles α and β , respectively) with orthogonal directions (x and y) tilt magnetization from the longitudinal B_0 direction to the x,y-plane and form a magnetization phase in the x,y-plane. This phase depends on the flip angles α and β and, therefore, can be used for inferring information on B_1 field. However, the measured phase of MR signal deviates from the magnetization phase—it depends also on different instrumental factors (RF coil mutual locations, cable length, etc.) (40). To eliminate these effects, Morell (35) proposed to use the second measurement with another pair of RF pulses (α , $-\beta$), and then to calculate the difference between the phases in these 2 measurements (in the original paper by Morell [35], the pairs $[2\alpha, \alpha]$ and $[2\alpha, -\alpha]$ were used). In Supporting Information (SI) to this paper, we analyze a choice of the flip angles (α , β). Our analysis is based on 2 main principles: first, the flip angles should be chosen in such a way to minimize an uncertainty in their determination from experimental data; second, experimental values of flip angles that can be used in human experiments are restricted by the RF power deposition characterized by the specific absorption rate. The preferences dictated by these 2 principles lead to different requirements for the optimal flip angles (see Supporting Fig. S1), and an optimal choice is a trade-off between these

requirements. As a result, we found (and used in our experiments) that the best choice corresponds to $\alpha = \beta \sim 90^\circ$.

The quantity ϕ corresponding to the difference of the MR signal phases generated by the pairs of RF pulses (α, α) and $(\alpha, -\alpha)$ can be presented as

$$\Delta\phi = \arctan[Q(\alpha, p)],$$

$$Q(\alpha, p) = \frac{2\Omega^2 \cdot (p^2 + \alpha^2 \cos\Omega)}{\alpha^2 \cdot (1 - \cos\Omega) \cdot (\alpha^2 + (p^2 + \Omega^2) \cdot \cos\Omega)}, \quad [1]$$

where $\omega = 2\pi \cdot f$ is the B_0 -related frequency shift, $p = \omega \cdot \tau$, $\Omega = (\alpha^2 + p^2)^{1/2}$ and τ is the duration of each RF pulse.

However, Equation [1] does not take into account magnetization relaxation effects and details of MRI pulse sequence used for data acquisition. These effects are considered below.

Effects of Relaxation on Signal Phase

To elucidate effects of relaxation properties, we need to solve the Bloch equations for magnetization $\mathbf{M}$, accounting for the RF field B_1 , the B_0 -related frequency shift ω and longitudinal and transverse relaxations described by R_1 and R_2^* relaxation rate constants, respectively. For our purpose, it is convenient to present the Bloch equations in the rotating frame in the following matrix form:

$$\frac{\partial \mathbf{M}}{\partial t} = \hat{\mathbf{A}}\mathbf{M} + \mathbf{P}, \quad \hat{\mathbf{A}} = \begin{pmatrix} -R_2^* & \Delta\omega & -\gamma B_{1y} \\ -\Delta\omega & -R_2^* & \gamma B_{1x} \\ \gamma B_{1y} & -\gamma B_{1x} & -R_1 \end{pmatrix}, \quad \mathbf{P} = (0, 0, R_1 \cdot M_0)^T, \quad [2]$$

where M_0 is the nominal magnetization and γ is the gyromagnetic ratio.

The solution of Equation [2] with the initial magnetization $\mathbf{M}(0)$ is

$$\mathbf{M}(t) = \exp(\hat{\mathbf{A}} \cdot t) \cdot (\mathbf{M}(0) - \mathbf{V}) + \mathbf{V}, \quad \mathbf{V} = -\hat{\mathbf{A}}^{-1} \cdot \mathbf{P}. \quad [3]$$

In what follows, we consider a sequence comprising a train of double-RF pulses. In each pair, the first RF α -pulse of duration τ_α and strength $B_{1x}(\alpha = \gamma \cdot B_{1x} \cdot \tau_\alpha)$ along the x-axis is immediately followed by the second RF β -pulse of duration τ_β and strength $B_{1y}(\beta = \gamma \cdot B_{1y} \cdot \tau_\beta)$ about the y-axis.

In the interval between the pairs of RF pulses, the longitudinal and transverse components of magnetization evolve as:

$$\mathbf{M}_{n+1}^{(-)} = \exp(\hat{\mathbf{A}}_0 \cdot (TR - \tau_\alpha - \tau_\beta)) \cdot (\mathbf{M}_n^{(+)} - \mathbf{V}_0) + \mathbf{V}_0, \quad [4]$$

where $\mathbf{M}_n^{(-)}$ and $\mathbf{M}_n^{(+)}$ are the magnetizations before and after the n -th pair of the double RF pulses, respectively (the index n numerates pairs of the double RF pulses in the train), the matrix $\hat{\mathbf{A}}_0$ is given by Equation [3] with $B_{1X,Y} = 0$, $\mathbf{V}_0 = -\hat{\mathbf{A}}_0^{-1} \cdot \mathbf{P}$, and TR is the repetition time. In what follows, we will only consider RF pulses with equal duration $\tau_\alpha = \tau_\beta = \tau$.

The magnetization $\mathbf{M}_n^{(+)}$ after the n -th pair of the double RF pulses can be presented as

$$\mathbf{M}_n^{(+)} = \exp(\hat{\mathbf{A}}_\beta \tau_\beta) \cdot \exp(\hat{\mathbf{A}}_\alpha \tau_\alpha) \cdot (\mathbf{M}_n^{(-)} - \mathbf{V}_\alpha) + \exp(\hat{\mathbf{A}}_\beta \tau_\beta) \cdot (\mathbf{V}_\alpha - \mathbf{V}_\beta) + \mathbf{V}_\beta, \quad \mathbf{V}_{\alpha,\beta} = -\hat{\mathbf{A}}_{\alpha,\beta}^{-1} \cdot \mathbf{P},$$

[5]

where the matrices $\hat{\mathbf{A}}_\alpha$ and $\hat{\mathbf{A}}_\beta$ are given by Equation [2] with $B_{1Y} = 0$ and $B_{1X} = 0$, respectively.

Although the matrix-form expression in Equation [5] for the magnetization $\mathbf{M}_n^{(+)}$ looks rather simple, the corresponding explicit expressions for $\mathbf{M}_n^{(+)}$ are very cumbersome and we do not provide them. However, Equations [4] and [5] can be readily used for obtaining analytical results under some simplifications and for numerical calculations in the framework of any matrix-operating software, (e.g., Wolfram “Mathematica,” Wolfram Research, Champaign, IL, and MATLAB, The MathWorks, Natick, MA).

In the brain, longitudinal and transverse relaxation times are ~ 1 s and ~ 20 – 50 ms, respectively, that is much longer than typical durations of RF pulses τ (< 1 ms). Therefore, for sufficiently short TRs, the following inequalities take place:

$$R_1 \cdot \tau \ll R_1 \cdot TR \ll 1, \quad R_2^* \cdot \tau \ll 1. \quad [6]$$

In this case, using Equations [4] and [5], it can be demonstrated that under steady-state condition and ideal spoiling (the transverse magnetization is completely annihilated before the RF pulses), in the linear approximation with respect to the small parameters $R_1 \cdot \tau$, $R_2^* \cdot \tau$, τ/TR , and $R_1 \cdot TR$, the difference of the phases for 2 pairs of the RF pulses (α , α) and (α , $-\alpha$) can be presented as

$$\Delta\phi = \arctan\left[Q(\alpha, p) + \left(\frac{\tau}{TR}\right) \cdot Q_1(\alpha, p) + (R_2^* \cdot \tau) \cdot Q_2(\alpha, p)\right], \quad [7]$$

where the function $Q(\alpha, p)$ is given in Equation [1]. Importantly, in this approximation, the relaxation rate constant R_1 does not enter Equation [7] (this constant appears only in the second-order terms with respect to the small parameters mentioned above). The parameter $(R_2^* \cdot TR)$, which is not small, does not enter Equation [7] because of the assumption of ideal spoiling (this issue will be addressed below). Note, however, that this equation should be modified for longer $TR \sim T_1$. The expressions for the function $Q_{1,2}(\alpha, p)$ for an arbitrary value of $p = \Delta\omega \cdot \tau$ are given in Equations [A1] and [A2] in Appendix.

Effects of Imperfect RF Spoiling on Signal Phase

All the above considerations were based on the assumption of a perfect spoiling of the transverse magnetization in each TR cycle. A standard and broadly used RF spoiling approach (41) comprises a RF phase cycling scheme (each RF pulse is accompanied by a phase shift $\Psi_{k+1} = \Psi_k + k \cdot \Psi_0$, where $\Psi_0 = 117^\circ$, and k is the RF pulse number) and an additional spoiling gradient preceding the next RF pulse. However, computer simulations (see below) revealed that for double RF pulses used in our approach, the effects of this standard RF spoiling turns out to be not sufficient to eliminate the transverse magnetization before the next RF pulse, which may lead to artifacts in B_1 mapping.

A development of a new RF spoiling scheme applicable to double RF pulses is beyond the scope of the present paper. Instead, in this study, we propose another approach based on introducing a correction to the expression for the phase difference ϕ , Equation [7], accounting for incomplete spoiling of the transverse magnetization.

To examine the effect of RF spoiling of the transverse magnetization on signal phase in our approach, we used computer simulations. For this purpose, using Equations [4] and [5], the Bloch equations were numerically solved for the train of consecutive TR cycles. To simulate the effect of the spoiling gradient, 50 isochromats were included in the simulations. An additional phase assigned to each isochromat was added to the isochromat's phase at the end of each TR cycle to simulate effects of spoiling gradient. These additional phases were uniformly distributed over the interval $[0, 2\pi]$ among isochromats. By the end of the simulation, the magnetization of the 50 isochromats were averaged. For $TR = 40$ ms and relaxation parameters typical for gray matter ($T_1 = 1200$ ms, $T_2^* = 70$ ms) and white matter ($T_1 = 800$ ms, $T_2^* = 50$ ms), the magnetization was shown to reach the steady-state within 60 steps, respectively. For CSF ($T_1 = 4000$ ms, $T_2^* = 400$ ms), ~250 steps are required.

The simulations were done for 2 cases: (1) perfect spoiling (ps): transverse magnetizations were artificially set to zero before each pair of RF pulses, and (2) standard experimental RF spoiling: using the scheme described above. All simulations were done in MATLAB (The

MathWorks) for equal durations and flip angles in the RF pulse pair ($\alpha = \beta = 90^\circ$). The results of simulations and their analysis are illustrated in Figure 1.

Obviously, for sufficiently long repetition times compared to T_2^* relaxation time constant, ($TR > T_2^*$), the transverse magnetization before the next RF pulse becomes negligible, and no spoiler is required. However, for shorter TR, φ_{exp} deviates from φ_{ps} at non-zero frequency shifts f (see Fig. 1). As demonstrated in Figures 1a and 1b, the difference $\delta\varphi = \varphi_{\text{exp}} - \varphi_{\text{ps}}$ is practically independent of the longitudinal relaxation time T_1 , but sensitive to the transverse relaxation time T_2^* . The dependence of $\delta\varphi$ on the frequency shift f is, generally, rather complicated (Fig. 1a). However, in brain applications, frequency shifts caused by the macroscopic field inhomogeneity usually do not exceed ± 300 Hz because of the good shimming capability of modern MRI scanners (42). In this interval, the simulations show that for the relaxation rate constant R_2^* typical for brain tissues the difference $\delta\varphi$ depends mainly on the dimensionless parameters $R_2^* \cdot TR$ and $p = 2\pi \cdot \Delta f \cdot \tau$ and can be well approximated by the following empirical expression:

$$\Delta\varphi_{\text{exp}} - \Delta\varphi_{\text{ps}} = \tanh(\kappa \cdot p), \quad [8]$$

where

$$\kappa = 2.38 \cdot \exp[-2.83 \cdot (R_2^* \cdot TR)^{0.65}] \quad [9]$$

It should be underlined that this expression is not universal and valid only in the intervals, $|\Delta f| \cdot \tau < 0.15$, $0.2 < R_2^* \cdot TR < 5.5$.

Taking into account Equation [8], Equation [7] can be modified as follows:

$$\Delta\varphi_{\text{exp}} = \arctan\left[Q(\alpha, p) + \left(\frac{\tau}{TR}\right) \cdot Q_1(\alpha, p) + (R_2^* \cdot \tau) \cdot Q_2(\alpha, p)\right] + \tanh(\kappa \cdot p), \quad (10)$$

where the functions $Q(\alpha, p)$ and $Q_{1,2}(\alpha, p)$ are given in Equation [1] and Equations [14] and [15], respectively. This expression can be used to account for the imperfect spoiling effect.

The importance of this correction is demonstrated in Figure 2, where the dependences of the parameter q (the ratio of the measured and input flip angles) on TR and f are shown for the values of T_1 and T_2^* characteristic for WM ($T_1 = 800$ ms, $T_2^* = 50$ ms), GM ($T_1 = 1200$ ms, $T_2^* = 70$ ms), and CSF ($T_1 = 4000$ ms, $T_2^* = 400$ ms). The top panels illustrate these dependencies for the results calculated from Equation [10] without the last term, whereas the lower panels illustrate the results obtained from Equation [10] with the last term.

It should be emphasized that the phenomenological expression for the correction term in Equation [10] is derived for the specific value of the angle, $\Psi_0 = 117^\circ$. However, different

scanner manufacturers use different values of this parameter, for example, $\Psi_0 = 117^\circ$, (VARIAN), $\Psi_0 = 50^\circ$ (SIEMENS), and $\Psi_0 = 115.4^\circ$ (GE). Although we cannot provide a general expression for the correction factor for arbitrary RF spoiling scheme, the analysis of the cases with $\Psi_0 = 50^\circ$ and $\Psi_0 = 115.4^\circ$ revealed that they still can be described in terms of Equation [10] but with the slightly different dependence of the coefficient κ on the product, $R_2^* \times TR$:

$$\begin{aligned} \Psi_0 = 50^\circ: \kappa &= 7.53 \cdot \exp[-3.99 \cdot (R_2^* \cdot TR)^{0.46}] \\ \Psi_0 = 115.4^\circ: \kappa &= 3.95 \cdot \exp[-3.27 \cdot (R_2^* \cdot TR)^{0.49}]. \end{aligned} \quad [11]$$

The correction factors for other Ψ_0 can be derived on a case-by-case basis in a manner similar to described above (the authors will be happy to assist for any other specific value of Ψ_0).

METHODS

MR Experiment

All experiments were performed on 3T SIEMENS TRIO (VB17) MRI scanner. In our experimental settings, we used 2 orthogonal rectangular RF pulses (flip angle $\alpha = 90^\circ$) with total duration of 800 μ s and a 20 μ s gap in between allowing for RF phase shift. The RF duration was selected as the shortest under the specific absorption rate limitation (number of echoes = 6, $TE_1 = 2$ ms, and $TE = 4$ ms). A navigator echo was collected during each TR to correct for artifacts induced by physiological fluctuation, as described previously (43). In all scans, the body coil was used as a transmitter. A 3D multi-gradient recalled echo (GRE) sequence with a combined non-selective RF excitation pulse consisted of the $(\alpha_x, \alpha_{\pm y})$ pairs was used as depicted in Figure 3.

Phantoms were prepared by adding 3.75 g of $\text{NiSO}_4 \times 6\text{H}_2\text{O}$ and 5 g NaCl per 1L of water. Experiments were run on: (1) a small spherical phantom with a radius of 1.5 cm using a 15-channel knee coil and image resolution of $1 \times 1 \times 1$ mm³; (2) a spherical phantom with a radius of 2.5 cm using a 12-channel head coil and image resolution of $4 \times 4 \times 4$ mm³; and (3) a healthy volunteer using a 32-channel head coil and image resolution of $2 \times 2 \times 2$ mm³. Human studies were approved by the local Institutional Review Board. The healthy volunteer provided informed consent.

For phantom experiments, the frequency shifts were introduced by (1) changing the B_0 frequency shift, and (2) changing the shimming profile along 1 phaseencoding direction. Two different TRs (30 ms and 200 ms) were used to compare the RF spoiling effect. To minimize effects resulting from the presence of voxels with coexisting fat and water compartments in human experiments, an additional Gaussian-type fat-suppression RF pulse preceding the phase-sensitive RF excitation pulse was added (bandwidth, 375 Hz; duration, 5120 μ s). With TR = 40 ms, a resolution of $2 \times 2 \times 2$ mm³ and GRAPPA (44) algorithm (acceleration factor of 2 and 24 autocalibrating lines in each phase encoding direction), each

scan on the healthy volunteer required 1 min 40 s to cover the whole brain and the upper spinal cord.

To demonstrate the B_1 effect on quantitative measurements of tissue relaxation parameters, we have also collected simultaneous multi-angular relaxometry of tissue (SMART) MRI data (22) with multiple flip angles of 5° , 10° , 20° , 40° , and 60° . This approach is based on (1) collecting GRE MRI data with multiple gradient echoes (we used gradient echo times $TE = 2.26, 6.19$, and 10.12 ms) and multiple flip angles (in this experiment we used 5° , 10° , 20° , 40° , and 60°), and (2) a new theoretical expression for GRE signal that takes into account the presence in the tissue of “free-” and “bound-water” pools (see details in (22)). By using a high resolution of $1 \times 1 \times 1 \text{ mm}^3$, $TR = 18$ ms and GRAPPA, the total acquisition time of the SMART sequence was 17min 30 s. Images from B_1 scans were registered to SMART images using “flirt” (45,46) tool in “FSL.” SMART data was then processed using the same procedure as described in (22). The maps of the SMART parameters— R_{1app} (apparent longitudinal relaxation rate constant), R_{1f} (longitudinal relaxation rate constant of the free-water compartment), and k_{app} (apparent magnetization exchange rate constant)—were calculated with and with-out B_1 correction.

Data Analysis

Raw data were loaded into MATLAB (The MathWorks) for processing that required several steps.

1. Using a strategy developed in (47,48), data from different channels are combined for each voxel in a single data set $S(TE)$:

$$S(TE_n) = \frac{1}{M} \cdot \sum_{m=1}^M \lambda_m \cdot S_m^*(TE_1) \cdot S_m(TE_n) \quad , \quad [12]$$

$$\lambda_m = \frac{1}{M\sigma_m^2} \cdot \sum_{l=1}^M \sigma_l^2$$

where the sum is over all the channels (M), S^* denotes complex conjugate of S , λ_m are the weighting parameters, and σ_m are the noise amplitudes (the index corresponding to the voxel position is omitted for clarity). This algorithm allows for the optimal estimation of quantitative parameters, such as MR signal frequency and decay rate constants and also removes the initial phase incoherence between the channels.

2. The frequency shift ($\omega = 2\pi \cdot f$) and R_2^* maps are calculated by fitting a mono-exponential function of echo time to the signal (generated from Equation [12]).
3. To get the phase difference ϕ_{exp} , data from different channels and different scans are combined for each voxel according to Equation [13]:

$$\Delta\varphi_{\text{exp}} = \arg \left[\frac{1}{M} \cdot \sum_{n=1}^N \sum_{m=1}^M \lambda_m \cdot S_m^{(1)}(TE_n) \cdot S_m^{(2)*}(TE_n) \right], \quad [13]$$

where the sum is over all channels and all echoes (N). The superscripts (1) and (2) correspond to the first and the second scans, respectively. This procedure is similar to the previously proposed by Bernstein et al. (40) for phase contrast MR angiography. It removes the initial phase φ_0 in the channels and increases the signal-to-noise ratio (SNR) in the φ image.

4. The measured flip angles α_{exper} are calculated by solving Equation [10] (using the results of steps 2 and 3) on the voxel-by-voxel basis. The results of flip angle measurements are represented by a parameter $q = \alpha_{\text{exper}} / \alpha_{\text{input}}$.

RESULTS

The accuracy of our method was first tested on a small phantom (1.5 cm in radius) that was placed at the isocenter of the scanner. In this phantom, that had a size much smaller than the characteristic size of RF transmit coil (body coil), no significant inhomogeneities were found (see Fig. 4): the parameter q was equal to 1.00 ± 0.03 (mean $\pm$ STD). This result provides experimental validation of our method as it shows an excellent agreement with the result obtained by the standard SIEMENS RF transmitter calibration procedure (49–51).

The theoretical results for different frequency shifts were tested on a spherical phantom (2.5 cm in radius). As shown in Figure 5 (lower panels), no imperfect spoiling effects were observed for long TR (200 ms). However, when TR was reduced to 30 ms, the estimated flip angles α_{exper} (and corresponding histograms of the parameter q) obtained by using Equation [1] were shifted in the direction depending on the sign of the frequency shift f , that associated with the imperfect spoiling of the transverse magnetization. However, when α_{exper} was calculated from Equation [10], the q -maps and corresponding histograms become practically identical with the maps and histograms obtained for long TR, as shown in Figures 5 and 6, respectively.

The validity of our correction approach has also been demonstrated in another experiment in which inhomogeneous frequency shift was introduced by changing the shimming profile along 1 phase-encoding direction (see Fig-7).

Note that before correction, the q map is strongly asymmetric along the phase-encoding direction and deviates from the result obtained for long TR when RF spoiling does not create artifacts. After correction, results for short TR practically coincide with the result for long TR.

The maps of the SMART parameters— R_{1app} (apparent longitudinal relaxation rate constant), R_{1f} (longitudinal relaxation rate constant of free-water compartment), and k_{app} (apparent magnetization exchange rate constant)—calculated without and with correction are shown in Figure 8. The corresponding histograms of the SMART parameters are shown in Figure 9

along with the difference maps (before and after B_1 correction). The B_1 correction of the flip angles (i.e., accounting for the deviation of the actual flip angles from their input scanner values) changes the values of the SMART parameters by up to 50%, 85%, and 50% for R_{1app} , R_{1f} , and k_{app} , respectively. With the strongest deviations in the areas of the large differences between assigned and actual flip angles (compare images in Fig. 9 with B_1 map in Fig. 8)

DISCUSSION

In this study, we focused on the effects of imperfect RF spoiling and relaxation on the phase-based B_1 mapping technique with 2 orthogonal excitation RF pulses used for encoding B_1 information in the signal phase. We demonstrated that the MR signal phase generated by the pair of successive orthogonal RF pulses depends not only on B_1 field strength but on the MR signal relaxation properties (T_1 and T_2^*) and (most importantly) on imperfections of RF spoiling as well. These effects have not been considered previously. To account for these effects, we proposed to append a GRE echo train with multiple gradient echoes to the double RF pulse signal excitation. The gradient echo train of the GRE sequence allows mapping the local frequency shifts and transverse relaxation of the MR signal. Combining the phase, frequency and R_2^* relaxation time constant measurements provided by the proposed sequence with the theoretical expression derived for the MR signal phase, makes it possible to calculate the local B_1 field strength. Because the MR signal phase depends on instrumental factors and is diverse across different RF channels in the phased-array coil, we used the previously developed optimized methods for combining multi-channel data (40,48), allowing for optimal parameters' estimation.

Our computer simulation revealed that in the presence of B_0 -related frequency shifts, the standard RF spoiling procedure does not completely destroy the transverse magnetization excited by the pair of orthogonal RF pulses. Such incomplete spoiling substantially affects the accuracy of B_1 mapping (if not accounted for) when using short TRs. As shown in Figure 2 (upper row), in the presence of frequency shifts Δf without accounting for RF spoiling artifact, the calculated q -values (the ratio of measured and input flip angles) for all 3 tissue types dramatically deviate from unity. The deviations are more severe for short TRs and higher frequency shifts. In this study, we propose a new method that allows B_1 mapping even in the presence of imperfect RF spoiling. As shown in Figure 2 (bottom row), after applying the new approach the calculated q -values become very close to 1.

As expected, for long TRs (>100 ms), the imperfect spoiling effect is negligible for GM and WM. However, for short TRs, without correction, the parameter q substantially deviates from 1, especially for CSF that has much longer T_1 and T_2^* than WM and GM. The lower panels in Figure 2 clearly show that the correction procedure works well for WM and GM. Even for CSF, in which the imperfect spoiling effect is more severe than in other tissues because of its longer T_1 and T_2^* the correction procedure substantially decreases the deviation of the parameter q from 1: $|q - 1| < 0.1$.

The new method was tested on spherical phantoms (Figs. [4 and 5], and 7). As described above, the computer simulation results have shown that frequency shifts, imperfect spoiling of the transverse magnetization and relaxation had significant adverse effect on B_1 mapping results when using double RF pulses and short TRs. These effects were also revealed in the phantom results. As shown in Figures 5 and 7, the calculated q -values were severely distorted in the regions with high frequency shifts. After corrections, the q -values became consistent with those acquired with longer TRs, for which the effect of imperfect RF spoiling was minimal. For human studies, significant changes of SMART parameters were observed after applying corrections, which indicated that the correction accounting for imperfect RF spoiling and relaxation is crucial for phase-based B_1 mapping techniques using double RF pulses.

In our study, we used a phased-array RF receive coil to map the B_1 field generated by the transmit coil in each imaging voxel. To optimize this measurement, the signals from different channels of the phased-array were combined according to Equations [12] and [13], as proposed in (47,48). However, our approach, based on using 2 orthogonal excitation RF pulses, can also be used for mapping the field from each individual channel in transmit-receive coils. This is important for more accurate implementations of parallel transmission approaches (1–3).

Our approach has certain restrictions because it relies on an assumption that the system under consideration can be described in terms of a single compartment model (or a multi-compartment model where all tissue compartments in the voxel have the same MR frequency). This, however, does not create a problem for most biological tissues possessing sufficiently small frequency differences between the compartments. However, the situation is substantially different in the regions containing both the fat and the water components because the frequency shift between them is high enough ($\sim 430\text{Hz}$ at 3T). An accurate quantification of B_1 in such areas requires an approach allowing for separation of the fat-water components or suppression of one of them. In our experiments, we use a Gaussian type fat-suppression RF pulse preceding the phase sensitive RF excitation pulse to minimize this problem. The current studies use non-selective RF pulses, which restricts this method to 3D applications. A slice-selective gradient in 2D applications could cause large frequency shifts, therefore bringing in additional artifacts.

CONCLUSION

In this study, we proposed and validated a phase-based mapping method that can be used to get fast, accurate B_1 maps using multi-channel phased-array coil. Our theoretical consideration includes effects of MR signal longitudinal and transverse relaxation that were not considered in previous studies. We have also demonstrated that the standard RF spoiling is not sufficient for the sequence with 2 orthogonal RF pulses and might create systematic errors in the B_1 mapping if not accounted for. To deal with this adverse effect, we proposed a new method that allows substantially improve B_1 estimation even for short TR. We also demonstrated that the effect of imperfect RF spoiling becomes negligible for TR longer than 200 ms for human brain at 3T. The feasibility of this method is demonstrated on phantoms and a healthy volunteer.

APPENDIX

The functions $Q_{1;2}(\alpha, \beta)$ entering Equation [7]

$$Q_1(\alpha, p) = \frac{N_1(\alpha, p)}{D_1(\alpha, p)} \quad [A1]$$

$$N_1(\alpha, p) = 2 \cdot \left\{ p^2 \Omega \cdot (3\alpha^4 + 10\alpha^2 p^2 + 4p^4) + \right.$$

$$+ 4p^2 \cos \Omega \cdot \left[\Omega \cdot (2\alpha^4 + 3\alpha^2 p^2 + 2p^4) - \alpha^4 \sin \Omega \right] +$$

$$\left. + (\alpha^2 + 2p^2) \cdot \left[(3\alpha^4 - 4\alpha^2 p^2 - 4p^4) \cdot \sin \Omega + \alpha^2 \cos 2\Omega \cdot (p^2 \Omega + \alpha^2 \sin \Omega) \right] \right\}$$

$$D_1(\alpha, p) = (1 - \cos \Omega) \cdot \alpha^2 \Omega \cdot \left[\alpha^2 + (\alpha^2 + 2p^2) \cos \Omega \right]^2$$

$$Q_2(\alpha, p) = \frac{N_2(\alpha, p)}{D_2(\alpha, p)}$$

$$N_2(\alpha, p) = \alpha^2 \Omega \cdot (\alpha^4 - 4p^4) \cdot \cos \Omega \cdot \left[7\alpha^4 + 100\alpha^2 p^2 + 8p^4 + 2(3\alpha^4 + 8\alpha^2 p^2 + 4p^4) \cdot \cos \right.$$

$$\left. \Omega \right] +$$

$$+ \alpha^4 (\alpha^2 - 2p^2) (\alpha^2 + 2p^2)^2 \Omega \cdot \cos 2\Omega - 4\alpha^4 (\alpha^2 + 4p^2) (3\alpha^2 + 4p^2) \cdot \Omega^4 \cdot \cot \frac{\Omega}{2} -$$

$$- \frac{1}{2} \alpha^2 (\alpha^2 - 2p^2) (7\alpha^6 + 12\alpha^4 p^2 - 8p^6) \cdot \Omega \cdot \csc^2 \frac{\Omega}{2} + 2(2\alpha^4 - 3\alpha^2 p^2 + 2p^4) \cdot \Omega^7$$

$$\cdot \csc^4 \frac{\Omega}{2} +$$

$$+ 2\alpha^2 (3\alpha^8 + 8\alpha^6 p^2 + 14\alpha^4 p^4 + 24\alpha^2 p^6 + 24p^8) \cdot \sin \Omega +$$

$$+ 2\alpha^2 (\alpha^2 + 7p^2) \cdot \Omega^6 \cdot \csc^4 \frac{\Omega}{2} \cdot \sin \Omega + \alpha^4 (\alpha^6 - 108\alpha^4 p^2 - 12\alpha^2 p^4 + 8p^6) \cdot \sin 2\Omega \left\{ \right.$$

$$D_2(\alpha, p) = 2 \Omega (-1 + \cos \Omega) \cdot \alpha^4 \cdot \left[\alpha^2 + (\alpha^2 + 2p^2) \cos \Omega \right]^3$$

[A2]

$$Q_1(\alpha, 0) = \frac{2 \cdot (3 + \cos 2\alpha)}{\alpha \cdot \sin \alpha \cdot (1 + \cos \alpha)} \quad [16]$$

$$Q_1(\alpha, 0) = \frac{(3 + \cos 2\alpha) \cdot (1 + \alpha \cdot \cot \alpha)}{2\alpha \cdot \sin^3 \alpha}$$

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

ACKNOWLEDGMENT

The authors are grateful to Dr. Yulin Chang for providing a custom program for GRAPPA reconstruction. Jie Wen is a former fellow of National Multiple Sclerosis Society.

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FIG. 1.

Effect of imperfect RF spoiling on signal phase. **(a)** The difference between φ simulated with the standard RF spoiling with $\varphi_0 = 117^\circ(\Delta\varphi_{\text{exp}})$ and that simulated with the perfect spoiling (φ_{ps}) as a function of the frequency shifts f for different T_1 ($T_2^* = 60$ ms, $TR = 40$ ms $\tau = 0.4$ ms). **(b)** The simulated difference $\varphi_{\text{exp}} - \varphi_{\text{ps}}$ as a function of the frequency shifts f for different T_2^* (symbols) ($T_1 = 1$ s, $TR = 40$ ms, $\tau = 0.4$ ms); the lines, fitting by the function in Equation [8]. **(c)** The fitting parameter κ (appearing in Equation [8]) as a function of the dimensionless product, ($R_2^* \cdot TR$) for different RF pulse duration T (symbols); the line, fitting by the function in Equation [8].

FIG. 2.

Maps of the parameter q (the ratio of the measured and input flip angles) as a function of Δf and TR before (top) and after (bottom) correction (accounting for imperfection of RF spoiling by means of the last term in [Eq. 10]). Simulations were performed for 3 sets of parameters typical for different tissue types: WM ($T_1 = 800$ ms, $T_2^* = 50$ ms), GM, ($T_1 = 1200$ ms, $T_2^* = 70$ ms), and CSF, $T_1 = 4000$ ms, $T_2^* = 400$ ms).

FIG. 3.

Schematic diagram of the pulse sequence. RF, RO, and PE represent radio frequency, read-out, and phase encoding, respectively. As in a standard RF spoiling procedure, for each subsequent TR, the phase of each RF pulse in the (α , α)-pair and the phase of receiver are changed by the increments of $\Delta\Psi_k = k \cdot \Psi_0$, where k is the phase encoding step number and Ψ_0 is a hardware-based parameter (in our Siemens scanner $\Psi_0 = 50^\circ$). The dephasing gradient is applied to destroy residual signal at the end of each TR block.

FIG. 4.

Map and histogram of the parameter q obtained from a small spherical phantom (1.5 cm in radius) using the sequence with $TR = 27$ ms. The histogram shows very narrow distribution with $q = 1.00 \pm 0.03$ (mean $\pm$ STD) therefore validating the accuracy of our approach against a gold standard RF calibration.

FIG. 5.

Maps of the parameter q obtained from a spherical phantom (2.5 cm in radius) with 2 different TRs (30 and 200 ms) and B_0 -related frequency shifts (0 and ± 200 Hz). For short TR (30 ms), q maps (3 left columns) were calculated without (top) and with (middle) correction according to Equation [10] with the correction factor in Equation [11] for our Siemens scanner ($\Psi_0 = 50^\circ$).

FIG. 6.

Histograms of the parameter q for the same experimental parameters as in Figure 5. **(a)** TR = 30 ms (before correction). **(b)** TR = 30 ms (after correction). **(c)** TR = 200 ms. Red, $f = -200$ Hz; green, $f = 0$; blue, $f = +200$ Hz.

FIG. 7.

Results obtained from a spherical phantom (2.5cm in radius) with different TRs (200 and 30 ms). Frequency shifts were introduced by changing the shimming profile along 1 phase-encoding direction. Calculated Δf and q maps are shown on the left. Histograms of q with different TR are shown on the right panel (green, TR = 200 ms; red, TR = 30 ms before correction; blue, TR = 30 ms after correction).

FIG. 8.

Comparison of SMART parameters ($R1_{app}$, $R1_f$ and k_{app}) maps without (second row) and with correction (third row) for imperfect RF spoiling in B_1 mapping. Top row: $R2^*$, frequency shift (Δf in Hz) and q maps. Units for $R2^*$, $R1_{app}$, $R1_f$ and k_{app} colorbars are s^{-1} . $R1_{app}$: apparent longitudinal relaxation rate constant; $R1_f$: longitudinal relaxation rate constant of the free-water compartment; k_{app} : apparent magnetization exchange rate constant (see details in (22)).

FIG. 9.

The difference maps and comparison of the histograms of SMART parameters (R_{1app} , R_{1f} and k_{app}) without (dashed lines) and with correction (solid lines) for imperfect RF spoiling in B_1 mapping. The difference maps were calculated by subtracting the SMART parameters before correction from the corresponding parameters after correction. After correction, the SMART parameters become more uniform across the entire brain, compared to those images before corrections. The biggest changes are in the brain areas with strongest actual B_1 deviation from the assigned B_1 value. The contrast between WM and GM also becomes higher. The SMART maps are shown in Figure 8 above.