Published in final edited form as:

Magn Reson Med. 2014 November ; 72(5): 1302–1310. doi:10.1002/mrm.25049.

B_1 Estimation using Adiabatic Refocusing: BEAR

Kalina V. Jordanova, Dwight G. Nishimura, and Adam B. Kerr

Magnetic Resonance Systems Research Laboratory, Department of Electrical Engineering, Stanford University, Stanford, California.

Abstract

Purpose—Accurate measurement of the non-uniform transmit radiofrequency field is useful for many applications in MRI, such as calibrating the scanner transmit system, evaluating coil performance, and improving image quality and quantitation. The radiofrequency field excitation amplitude (B_1) is often obtained by acquiring a B_1 map. In this study, a new B_1 mapping method is proposed.

Theory and Methods—The use of two adiabatic full passage pulses with different magnitudes applied as successive refocusing pulses results in a linear relationship between phase and B_1 field strength that is insensitive to the repetition time, off-resonance effects, T_1 , and T_2 . Using this method, B_1 mapping can be localized to a slice or 3D volume, with a spin-echo acquisition that is appropriate for fast projection measurements.

Results—This new method is shown to agree well with the Bloch-Siegert B_1 mapping method for both phantom and *in vivo* B_1 measurements at 1.5T, 3T and 7T. The method's ability to acquire accurate projection B_1 measurements is also demonstrated.

Conclusion—This method's high dynamic range, ability to make fast projection measurements, and linear quantitative relationship between phase and B_1 make it an ideal candidate for use in robust transmitter gain calibration.

Keywords

RF field; B_1 mapping; adiabatic pulses

Introduction

Accurate measurement of the non-uniform transmit radiofrequency (RF) field is useful for many applications in MRI; e.g., for calibrating the scanner transmit system, evaluating coil performance, designing parallel transmit excitation pulses (1, 2), and quantitative imaging. Typically, the measurement of the field in the form of a B_1 map is preferred over simulations, since the field non-uniformity is dependent on the scan subject. B_1 mapping is especially important for high main-field imaging ($B_0 = 3T$), where the variations in the RF field are large due to a reduction of the RF wavelength to the order of the typical sample size

Address correspondence to: Kalina Jordanova, Packard Electrical Engineering, Room 208, 350 Serra Mall, Stanford, CA 94305-9510, kalinaj@stanford.edu.

(3). There are a number of existing B_1 mapping methods that can be broadly separated into two categories: signal magnitude-based methods and signal phase-based methods.

Signal magnitude-based methods use the relationship between B_1 and the image magnitude to create a B_1 map. Many methods have been proposed for acquiring images so that their magnitude is sensitive to variations in B_1 . These include the use of progressively increasing ip angles (4), image signal ratios using different ip angles (5–7), different TRs (8), stimulated echoes (9,10), the pre-saturation effect (11,12), saturation recovery with different delay times (13), and signal null (14). The use of many of these methods is widespread; however, magnitude-based B_1 mapping can suffer from a variety of complications, including T_1 and T_2 dependence, long scan times, and a limited B_1 range over which the map can accurately be determined.

Signal phase-based methods use the relationship between B_1 and the image phase to create a B_1 map. These methods include the use of the Bloch-Seigert (BS) shift phenomenon (15), adiabatic pulses (16, 17), and other signal phase manipulation techniques (18, 19). Compared to magnitude-based methods, phase-based methods can offer some improvement. The method proposed by Morrell (18) offers an improved accuracy over a larger range of flip angles, but requires a long TR. The BS method (15) has been shown to be accurate for a large range of B_1 variations and is insensitive to T_1 and T_2 . However, care in implementation of the BS method must be taken when high variations in B_0 are expected, since it may be sensitive to these variations.

The use of adiabatic pulses offers alternative methods of phase-based B_1 mapping. These methods are insensitive to T_1 variations, but have their own drawbacks. One particular drawback is high SAR, which may make the application of these methods difficult at high magnetic field strengths. The method proposed by Hennel (17) is sensitive to B_0 variations, and requires phase unwrapping to create a B_1 map. The method proposed by Park and Garwood (16) eliminates the B_0 dependence, but has a large quadratic phase sensitivity that also requires phase unwrapping to be applied.

We describe a new phase-based method for B_1 mapping, which we have named BEAR: “ B_1 estimation using adiabatic refocusing”. This method uses adiabatic pulses to induce a spin-echo with phase that has a linear dependence on B_1 , and is insensitive to off-resonance, repetition time (TR), T_1 and T_2 . We discuss the theory behind the BEAR method, and examine the resulting signal phase dependence due to B_1 variations. We then validate BEAR’s performance in simulation and experimentally by comparison to the BS method. Finally, we discuss some of the benefits to using the BEAR method, especially for the transmit gain calibration of a scanner.

Theory

Basic Sequence

In a spin-echo sequence using a selective adiabatic full passage (AFP) pulse, the phase will vary quadratically over the slice profile, and with respect to B_1 variations (20). The second echo in a twice-refocused spin-echo sequence using two repeated identical AFP pulses will

have no phase variation over the slice profile (21, 22). The BEAR method relies on the observation that by changing the relative magnitude of the two AFP refocusing pulses, the phase of this echo will depend approximately linearly on B_1 while maintaining very little variation over the slice profile.

Figure 1 shows the BEAR sequence with two hyperbolic secant (HS1) (21) AFP pulses of peak B_1 magnitude $\delta B_{1,\text{nom}}$ and $B_{1,\text{nom}}$, where δ is a scaling factor and $B_{1,\text{nom}}$ is the nominal peak B_1 of the second AFP pulse. The equations governing the magnitude and frequency sweeps of the adiabatic pulses are

$$B_1(t) = B_1 \text{sech}(\beta t) \quad (1)$$

$$\Delta\omega_{RF}(t) = -\mu\beta \tanh(\beta t) \quad (2)$$

$$B_1^{\text{eff}}(t) = \sqrt{(\gamma B_1(t))^2 + (\Delta\omega_{RF}(t))^2} \quad (3)$$

where $\omega_{RF}(t)$ is the frequency offset of the pulse from resonance, and β and μ are parameters that determine the bandwidth and adiabatic threshold of the pulse. The adiabatic threshold is the minimum B_1 for which an AFP pulse will ensure refocusing of the transverse magnetization, by keeping the magnetization in spin-lock throughout the duration of the pulse (21). We define the adiabatic threshold, B_{1A} , as the minimum B_1 that ensures refocusing of 90% M_{xy} . As seen in Figure 1, imaging can be confined to a specified volume by making the refocusing pulses selective in a different direction than the excitation pulse.

Assuming any residual signal is effectively spoiled at the end of each repetition time, there should be no dependence on T_1 or TR in the phase of the acquired signal. Additionally, in order for spins to be contributing signal at the echo time, they must have T_2 values greater than the adiabatic pulse widths. This in turn indicates that T_2 relaxation can be ignored during the duration of the AFP pulses. Thus, we expect this method to be insensitive to T_1 , T_2 and TR effects.

B_1 Encoding

To determine the phase dependence on B_1 using the BEAR method, we can look at the accumulated phase at the end of the sequence shown in Figure 1.

Assuming spin-lock, the phase accrued due to one adiabatic pulse can be calculated by integrating the effective RF field (Equation 3) over the pulse length (20). The phase accrued for the BEAR sequence is then calculated by taking the difference of the phase accrued from each individual adiabatic pulse. The phase dependence at the end of one TR of the BEAR sequence is

$$\phi(x) \approx \int_0^T \left(\sqrt{(\gamma x B_1(t))^2 + (\Delta\omega_{RF}(t))^2} - \sqrt{(\gamma x \delta B_1(t))^2 + (\Delta\omega_{RF}(t))^2} \right) dt \quad (4)$$

Here, the variable x is included to represent the RF field variation, with the actual peak B_1 given by $B_1 = xB_{1,\text{nom}}$.

To eliminate phase effects unrelated to the B_1 field strength, the phase difference between two images is used to create a B_1 map. As indicated in Figure 1, the second image is acquired with the magnitudes of the adiabatic pulses switched temporally. Therefore, the resulting phase of the BEAR method is calculated by taking the difference of the phase accrued in each image, and will be twice that given in Equation 4.

While the analytic expression given in Equation 4 holds for spin-lock ($xB_{1,\text{nom}} > B_{1A}$), we avoid any errors introduced by this underlying assumption by instead using numerical Bloch simulations (23), neglecting T_1 and T_2 relaxation, to determine the signal phase dependence on B_1 for this sequence, and to construct a look-up table for mapping phase to B_1 . This was the method used for the results presented in this paper.

Alternatively, to facilitate implementation, it is possible to accurately model the phase dependence on $|B_1|$ with a first- or second-order polynomial. If Bloch simulations are not performed, this approximation can be used to map measured phase to B_1

$$B_1(\phi) = a_0 + a_1\phi + a_2\phi^2 \quad (5)$$

The coefficients a_0 , a_1 and a_2 for a set of adiabatic pulse parameters are given in Table 1. The coefficients were calculated via a polynomial fit of the Bloch simulation results to the larger B_1 range indicated in Table 1. The first sub-row for each δ shows a quadratic fit, while the second sub-row shows the results for a linear fit. The first-order polynomial fit is used to calculate the approximate phase sensitivity of the method, $S_{\text{BEAR}} = 1/a_1$. Included in Table 1 is the maximum error expected due to using the polynomial approximation rather than the numerical simulations, for specific ranges of B_1 .

As both the BS and BEAR methods are limited by SAR, the time efficiency of each method was calculated as the phase shift divided by the total energy of the relative pulses: the Fermi pulse for BS, and the adiabatic pulses for BEAR. (24).

Noise Analysis

To analyze the sensitivity to measurement noise of the BEAR method, we follow the methodology initially described for the BS method's noise analysis (25) to calculate the alpha-to-noise ratio (ANR) of the sequence

$$\text{ANR} = \frac{\alpha}{\sigma_\alpha} \quad (6)$$

Here α is the flip angle of the excitation pulse and σ_α is the noise standard deviation of the flip angle. The flip angle, α (radians), for the BEAR sequence can be written as a product of B_1 (G) and K_{EX} (radians/G), a constant that relates B_1 to flip angle for the excitation pulse of the sequence. Using Equation 5, if the quadratic component of the signal phase dependence

on B_1 is assumed to be negligible, the relationship between the flip angle and the measured BEAR phase, ϕ (radians), can be written as

$$\alpha = K_{EX} B_1 = K_{EX} (a_0 + a_1 \phi) \quad (7)$$

To find the standard deviation of the flip angle, we employ the law of error propagation, taking the partial derivative of Equation 7 with respect to ϕ , and use the substitution that for $SNR \gg 1$ the standard deviation of the measured phase is equal to the inverse of the magnitude SNR of the image, $\sigma_\phi = 1/SNR_{mag}$ (26). Thus

$$\sigma_\alpha = \sigma_\phi \frac{\partial \alpha}{\partial \phi} = \frac{K_{EX} a_1}{SNR_{mag}} \quad (8)$$

Combining Equations 6 – 8, the ANR of the BEAR sequence is

$$ANR = \frac{\alpha}{K_{EX} a_1} SNR_{mag} \quad (9)$$

Note that the ANR is proportional to the sensitivity of the method, $1/a_1$, and to the SNR of the image, as would be expected. Note that the sensitivity of the method increases with increasing ratio ($1/\delta$) of the refocusing pulse magnitudes, as can be observed from Table 1.

A B_1 -to-noise ratio (B_1NR) may also be useful for understanding the noise properties of the method, and more explicitly shows the dependence on B_1 . This can be determined directly from Equation 9 as

$$B_1NR = \frac{B_1}{a_1} SNR_{mag} \quad (10)$$

Note that if $B_1 > a_1$, the B_1NR of the method is greater than the magnitude SNR of the image. The value B_1/a_1 or B_1S_{BEAR} gives the relative increase of the B_1NR to SNR_{mag} for a particular B_1 . When $B_1 < B_{1A}$ the sensitivity varies as a function of B_1 and the refocused signal magnitude decreases. Then, the instantaneous sensitivity, S , can be calculated, and B_1S gives the relative increase of B_1NR over SNR_{mag} if the signal magnitude is unchanged. The value of $|M_{xy}|S/S_{BEAR}$ indicates how much the B_1NR degrades due to a decreased sensitivity and magnitude for a particular B_1 .

We have thus far examined the sensitivity and noise properties of the BEAR method. As opposed to other conventional B_1 mapping methods, BEAR has a refocused signal with a nearly linear phase dependence on B_1 that is insensitive to T_1 , T_2 , and off-resonance effects. Its phase sensitivity to variations in B_1 can be easily tuned by adjusting the parameter δ .

Methods

Adiabatic Pulse Parameters

The parameters of the AFP pulses ($T/\beta/\mu/\delta/B_{1,\text{nom}}$) were chosen with several properties of the BEAR sequence in mind.

To minimize the scan time and the effect of T_2 on the SNR, the pulse duration T was kept short. On the other hand, T could not be made arbitrarily short since the effective length of the pulses, a function of T and β , should be long to minimize truncation of the magnitude sweep near the beginning and end of the pulse (27).

The parameters β and μ affect the adiabatic threshold ($B_{1A} \gg \sqrt{\mu\beta}/\gamma$) and bandwidth ($BW = 2\mu\beta$) of the pulse (28). Increasing β and μ increases the bandwidth of the pulse, but also increases the adiabatic threshold. These parameters were chosen to keep the adiabatic threshold low, so that the sequence performs well over a large range of B_1 variations, while maintaining a reasonable bandwidth for the off-resonance expected from chemical shift and main-field inhomogeneity.

The expected range of variations in B_1 , along with the desired ANR, pose restrictions on δ and $B_{1,\text{nom}}$. Since the peak B_1 of the smaller adiabatic pulse is $\delta B_{1,\text{nom}}$, the adiabatic threshold of the sequence is effectively higher than the adiabatic threshold of the individual pulses by the factor $1/\delta$. The parameter δ is inversely related to the adiabatic threshold and to ANR, since a_1 increases with increasing δ (Table 1), resulting in a tradeoff between a low B_{1A} and a high ANR for a particular value of δ .

Projection measurements pose an additional restriction on δ . To ensure accurate projection B_1 measurements that closely match the average B_1 for a specified volume, phase wrapping must be eliminated completely. This limits the allowable total phase variation over the range of expected B_1 variations. For high field imaging, where the variations in B_1 are large, this may require an increased δ . Note that in general, increasing δ will decrease the ANR of the images. However, also note that in projection measurements, the image SNR increases because the effective voxel size increases. Therefore a δ can be chosen such that the sensitivity is low enough to ensure an accurate projection measurement, while keeping the overall ANR high depending on the size of the volume being imaged.

Keeping these criteria in mind, the AFP pulses were designed with parameters $T/\beta/\mu$ equal to 12 ms / 800 rad·s⁻¹ / 5.5, giving a bandwidth of 1.4 kHz, and truncation of the HS1 pulses such that $\text{sech}(\beta T/2) = 0.0165$. The adiabatic threshold, B_{1A} , for this sech pulse is 0.095 G, and the adiabatic threshold of the sequence is 0.095/ δ G. With a $B_{1,\text{nom}}$ of 0.20 G, $B_{1,\text{nom}} > B_{1A}/\delta$ for $\delta > 0.48$. A δ of 0.8 was chosen, allowing for a 41% variation in B_1 before the adiabatic threshold is reached.

Sequence Parameters

For both the BEAR sequence and the comparative BS sequence, data acquisition used a single-slice acquisition protocol on 1.5T, 3T and 7T GE scanners. A GE birdcage head coil was used for reception, and the GE body coil was used for transmission at 1.5T and 3T,

while the head coil was used for transmission at 7T. For *in vivo* scans, informed consent was acquired from all volunteer subjects, in accordance with institutional review board guidelines.

A tip angle of 42° was chosen to ensure a large signal from grey and white matter at 1.5T using the BEAR sequence, based on simulations of the steady-state signal. At different field strengths, this optimal tip angle will change due to changes in T_1 . However, for all scans conducted in this study the 42° tip angle was used. The matrix size of the images was $256_x \times 128_y$ pixels, with a FOV of 24 cm. A TR of 500 ms was chosen to limit the SAR at the higher field strengths, and used for all images. Shorter TRs were tested at 1.5T to ensure a lack of TR dependence for the BEAR method. The total scan time using a 500 ms TR was about 1.2 minutes per individual image, and 2.4 minutes per pair of images. A single signal acquisition was used per image. Note that the scan time can be decreased by using alternative readout methods, such as a spiral or EPI readout. The TE for BEAR with the AFP parameters previously mentioned was approximately 49 ms.

The BS method used an 8-ms Fermi pulse, with ± 4 kHz off-resonant frequency, resulting in a TE of approximately 17 ms. The same tip angle, matrix size, FOV, and TR as the BEAR method were used for the BS method. The crusher area for the BS method was set to induce 6 cycles of phase variation over the slice thickness.

To eliminate phase effects unrelated to the B_1 field strength, phase-difference images were made from multiple acquisitions. For BEAR, the second acquisition reversed the order of the two adiabatic pulses; for BS, the second acquisition negated the off-resonant frequency of the Fermi pulse. For $\delta = 0.8$ and $B_{1,nom} = 0.20$ G, the phase sensitivity was 38.1 rad/G. The BS method sensitivity was 57.7 rad/G, calculated by the tangent line to the curve at $B_1 = B_{1,nom}$.

To determine the accuracy of the simulated look-up tables used to map phase to B_1 for both methods, a sequence of scans was conducted at 1.5T using a small uniform phantom. For these scans only, the image size was 64×64 pixels and a TR of 100 ms was used. All other parameters were kept the same as above. For each scan, the transmit gain was manually decreased, and the resulting B_1 maps were compared to predicted B_1 values based on the transmit gain setting. The ratio $B_1/B_{1,max}$ was compared to the transmitter attenuation, where $B_{1,max}$ was calculated as the average measured B_1 using both the BEAR and BS measurements and a particular region of interest.

For the BEAR method, imaging could be confined to a specified volume by making the refocusing pulses selective in Y (Figure 1), and limiting the X readout receiver bandwidth. The limited volume size was 6 cm. Fast, 1D projections could also be acquired using a single readout with $k_y = 0$. For comparison to these fast projection acquisitions, 2D B_1 maps were also acquired, and their B_1 magnitude averaged along Y.

The spoiler and crusher sizes of the BEAR method were chosen to eliminate stimulated echoes and excitation from the transition bandwidth of the first adiabatic pulse. To avoid stimulated echoes, different sizes were chosen for each of the crushers, with a ratio of 1:2 between them, as well as for the spoiler, with a ratio of 1.67 to the smaller crusher.

Eliminating the unwanted excitation from the transition band of the first adiabatic pulse was especially important for volumetric imaging using the BEAR method.

Results

BEAR's signal phase dependence on peak B_1 and δ is shown in Figure 2a for numerical Bloch simulations and for the analytical expression given in Equation 4. The signal has approximately linear phase dependence on B_1 for $B_1 > B_{1A}$. The BS phase dependence on B_1 for an 8-ms 4-kHz-offset Fermi pulse is shown for reference.

Figure 2b shows the efficiency of the BS method used in this study, and for the BEAR method for $\delta = 0.7, 0.8, 0.9$. For $\delta = 0.7$ and 0.8 , the efficiency of the BEAR method is higher than the efficiency of the BS method for a large range of B_1 . BEAR's efficiency using $\delta = 0.9$ drops below the efficiency of the BS method.

The simulated magnitude and phase of the refocused M_{xy} , as a function of B_1 and off-resonance frequency, are shown in Figure 2c–d, illustrating BEAR's insensitivity to off-resonance over the effective bandwidth of the refocusing pulses. Figure 2c also shows signal dropout for $B_1 < B_{1A}$, which will ultimately result in decreased B_1 NR and an expected increase in B_1 map errors.

Another source of increased B_1 map error is due to the phase profile itself: the phase sensitivity decreases, and there is an increased phase variation across the slice profile with respect to off-resonance frequency. B_1 S, which determines the effect on B_1 NR due to a varying sensitivity, is shown in Figure 2e as a function of B_1 and for a range of δ s. For $\delta = 0.8$, B_1 S = [4.32, 1.23] rad at $B_1 = [B_{1A}, 0.5B_{1A}]$. Figure 2f shows the B_1 NR ratio, $|M_{xy}|S/S_{\text{BEAR}}$, which represents the fractional decrease of B_1 NR due to decreased phase sensitivity and signal magnitude. For $\delta = 0.8$, $|M_{xy}|S/S_{\text{BEAR}} = [0.868, 0.076]$ at $B_1 = [B_{1A}, 0.5B_{1A}]$.

Figure 2g shows the percent variation in the resulting B_1 map due to phase variation across the slice. This is calculated by first mapping the phase profile shown in Figure 2d to B_1 using the curve shown in Figure 2a for $\delta = 0.8$, and then finding the percent difference between the calculated on-resonance B_1 values from the B_1 values of the remainder of the slice profile. This results in minimum/maximum B_1 errors across the slice profile of [−0.592, 0.163]% at $B_1 = B_{1A}$, and [−0.935, 0.123]% at $B_1 = 0.5B_{1A}$.

Figures 3a–b show phantom results for which the transmit gain was manually varied for a series of scans. The mean and standard deviations for $B_1/B_{1,\text{max}}$ are shown for two different regions of interest as a function of the transmitter attenuation. In general, the mean values agree with the attenuation values, with relatively small standard deviations except at very low B_1 . Figure 3c shows the percent difference between the measured $B_1/B_{1,\text{max}}$ mean and the transmitter attenuation. For a wide B_1 range, even below the adiabatic threshold for the BEAR method, the error is well below 5%. For reference, Figure 3d shows example B_1 maps acquired using each method.

Spin-echo BEAR B_1 maps and gradient-echo BS B_1 maps were acquired from a spherical phantom and *in vivo*, to compare the performance of the two methods. Figure 4a shows a BS

B_1 map of the spherical phantom ($T_1 = 107$ ms) acquired with a TR of 500 ms on a 1.5T scanner. Figures 4b–c show BEAR B_1 maps of the same phantom acquired using TRs of 500 ms and 100 ms, demonstrating the method's insensitivity to TR. Figure 4d shows the percent difference between the BS B_1 map and the BEAR B_1 map using a TR of 500 ms, with a root mean square (RMS) difference of 0.0011 G between the BEAR method and the BS method.

Figures 4b,e–f show phantom BEAR maps as a function of δ , illustrating the effect of increasing ANR as δ is decreased. Figures 4b,g–h show phantom BEAR B_1 maps as a function of peak B_1 . These three maps were acquired with $B_{1,\text{nom}} = 0.20, 0.15$, and 0.10 G. The B_1 maps are scaled so that the nominal B_1 value corresponds to 0.20 G, for easier comparison. As the peak B_1 decreases, approaching the sequence adiabatic threshold of $B_{1A} = 0.119$ G with $\delta = 0.8$, the B_1 maps deviate from the expected output.

In vivo B_1 maps were acquired using the BEAR and BS methods at 1.5T, 3T and 7T, again with $B_{1,\text{nom}} = 0.20$ G. The results are shown in Figure 5. At lower field strengths, the BS and BEAR maps agree well, with an RMS of 0.0131 G between the two methods. At higher field strengths, the deviation between the two methods increases, with RMS differences for 3T and 7T of 0.0158 G and 0.0228 G respectively.

Figure 6 shows that B_1 projections acquired with BEAR are in agreement with computed projections of 2D B_1 magnitude maps at 1.5T and 3T. There exist slight deviations between the two, which are expected since the BEAR projection measurements are weighted by the signal magnitude. The average differences between measured and computed projections were 0.75% at 1.5T and 1.6% at 3T for scanning over a slice; 1.2% at 1.5T and 0.56% at 3T for scanning over a restricted volume. Thus, the BEAR method offers a fast way to acquire a reasonable approximation of the average B_1 using two projection acquisitions.

Discussion

The BEAR method is a new method of B_1 mapping that can be localized to a slice or 3D block volume with a spin-echo acquisition that is appropriate for fast projection measurements. This method uses adiabatic pulses to induce a phase that has a linear dependence on B_1 , and is insensitive to off-resonance and TR. As the method measures transverse magnetization phase perturbation, it is insensitive to T_1 and T_2 . The method has a large dynamic range as long as the AFP pulses operate over their adiabatic threshold. BEAR also provides good image quality even in regions of B_0 inhomogeneity due to its robust spin-echo acquisition, making it suitable for use in the transmit gain calibration of a scanner.

The sensitivity of the BEAR method can be easily tuned by varying the parameter δ . Furthermore, for the parameters shown in this study, the ANR is greater than the SNR of the underlying image magnitude for a large range in B_1 . As δ is decreased, the sensitivity, and ANR, will increase. However, as δ is decreased there will theoretically be an increased phase variation with respect to off-resonance frequency. The phase dependence on off-resonance frequency is also expected to increase as B_1 increases. In practice, for a B_1 range of up to 0.4 G, we see negligible effects of off-resonance frequency phase variations. In the case where the adiabatic pulses are non-selective and variations in the signal phase

sensitivity due to off-resonance frequency are unavoidable, an additional B_0 map can be acquired to correct for the varying phase sensitivity.

As the sensitivity of the method increases, phase unwrapping techniques may need to be employed, as in other phase-based B_1 mapping methods. This is especially true if large variations in B_1 are expected. A possible work-around is to use a larger δ to reduce the sensitivity to B_1 , which can provide an estimate of B_1 for a second map using a higher sensitivity (lower δ).

BEAR's B_1 NR for lower B_1 values depends on the adiabatic threshold of the AFP pulses used. This may affect the B_1 maps in regions with very large B_1 fluctuations, where the peak B_1 may fall below the adiabatic threshold in some portions of the image, resulting in a decrease in sensitivity and signal magnitude. We found through simulation that below the adiabatic threshold B_1 tends to be more highly underestimated than overestimated due to phase variations over the slice profile. Additionally, we found through scanned phantom measurements that the effect of decreased sensitivity and signal magnitude was below 5% for B_1 values ranging down to at least $0.5B_{1A}$. Nevertheless, there will be some error due to these effects at low values of B_1 . Thus, redesigning the adiabatic pulses for a lower adiabatic threshold, using a larger δ or $B_{1,nom}$, or implementing the sequence with different pulse shapes with intrinsically lower adiabatic thresholds, such as HSn pulses (29), will increase the range of B_1 for BEAR. The limited dynamic range will also limit the BEAR method's application for parallel transmit field mapping, where the RF field typically varies drastically. A possible solution could be to use an all-but-one mapping scheme (30), where the RF field over the region of interest could be maintained over the adiabatic threshold.

SAR is a concern for BEAR, especially at higher field strengths. A long repetition time, such as the one used in this study, can help limit SAR. We observed that the method is not sensitive to TR. The long TE of the BEAR method may cause signal reduction in regions of short T_2 . This reduced signal will lower the SNR, thus lowering the B_1 NR of the measurement, and may result in B_1 measurement errors, especially at higher field strengths where T_2 is typically shortened or in different organs with short T_2 . Comparing efficiency between the BEAR and BS methods used in this study shows that the BEAR method can be more efficient than the BS method, depending on the value of δ used. The efficiency of both methods will be affected by the signal magnitude, which for the BS method is a function of T_2^* , and for the BEAR method is a function of T_2 , and of the magnitude of the refocused signal. Although T_2 is larger than T_2^* , BEAR has a longer TE than the BS method, and thus the final efficiency will depend on the parameters used for both sequences, tissue parameters, and the expected B_1 range determining the refocused signal magnitude for BEAR. Because the BEAR method is more efficient than the BS method for some values of δ , we expect that multislice applications may be used with the BEAR method similarly as to the BS method, however we do expect SAR to be a constraint on multislice acquisition.

In vivo B_1 maps acquired using the BEAR and BS methods at 1.5T, 3T and 7T showed increased variations in the difference between maps acquired with each method at high field strengths. This increase is most likely due to two main factors. First, the BS method's sensitivity to main-field inhomogeneity causes erroneous B_1 measurements, especially

visible near the sinuses and base of the head. Note that using an optimized BS pulse for B_0 variations (24), as well as shortening the TE of the BS implementation could ameliorate this effect. The BEAR method's robustness to off-resonance effects is evident by the smoothness of the maps even in regions of high inhomogeneity, such as in the sinuses. A second cause of the deviation between the BEAR and BS maps at high field strengths is the large variation in B_1 . In regions of very low peak B_1 , the BEAR maps may be affected due to a loss of adiabaticity. Apart from the regions of high B_0 inhomogeneity and very low B_1 , the BEAR and BS methods agree well, with RMS differences of 0.0131 G (1.5T), 0.0158 G (3T) and 0.0228 G (7T).

The BEAR method offers a direct quantitative relationship between phase and B_1 . This linear relationship allows for accurate estimation of the average B_1 magnitude in a limited volume by acquiring projection measurements. Given a perfectly homogenous phantom and uniform receive sensitivity, the projection measurements will exactly match the average B_1 ; however, due to signal non-uniformity some variations between the two are expected due to signal magnitude weighting of the projections. The ability of the BEAR method to restrict imaging to limited volumes may result in more accurate projection measurements, since in limited volumes where the signal is approximately uniform the projection measurements are expected to be quite accurate.

Projection measurements acquired at 1.5T and 3T were in agreement with computed projections of 2D B_1 magnitude maps. Projection measurements could not be acquired at 7T due to the use of a phased-array receive coil for data acquisition. The phase variation intrinsic to this multi-channel coil resulted in inaccurate B_1 projection measurements. A quadrature head coil could be used to acquire more accurate 7T projection measurements.

Conclusion

We have presented a new method for B_1 mapping using adiabatic pulses, named BEAR, and demonstrated its performance *in vivo* at 1.5T, 3T, and 7T. This method acquires accurate B_1 maps localized to a slice or 3D block volume with a spin-echo acquisition that is appropriate for fast projection measurements. BEAR's ability to tune to a particular B_1 range, insensitivity to B_0 , TR, T_1 , and T_2 , fast projection measurement acquisition, and linear quantitative relationship between phase and B_1 make it an ideal candidate for use in robust transmitter gain calibration.

Acknowledgments

We thank Mohammed Mehdi Khalighi and Brian Rutt for many helpful discussions and support in testing this method on a 7T high-field system, as well as Bob Dougherty for access to a 3T field system. This material is based upon work supported by the National Science Foundation under Grant No. DGE-1147470, by the National Institutes of Health under R01EB008108, and by GE Healthcare.

References

1. Katscher U, Börnert P, Leussler C, van den Brink JS. Transmit SENSE. *Magn Reson Med*. 2003; 49:144–150. [PubMed: 12509830]
2. Zhu Y. Parallel excitation with an array of transmit coils. *Magn Reson Med*. 2004; 51:775–784. [PubMed: 15065251]

3. Vaughan JT, Garwood M, Collins CM, Liu W, DelaBarre L, Adriany G, Andersen P, Merkle H, Goebel R, Smith MB, Ugurbil K. 7T vs. 4T: RF power, homogeneity, and signal-to-noise comparison in head images. *Magn Reson Med*. 2001; 46:24–30. [PubMed: 11443707]
4. Hornak JP, Szumowski J, Bryant RG. Magnetic field mapping. *Magn Reson Med*. 1988; 6:158–163. [PubMed: 3367773]
5. Insko EK, Bolinger L. Mapping of the radiofrequency field. *J Magn Reson A*. 1993; 103:82–85.
6. Stollberger R, Wach P. Imaging of the active B1 field in vivo. *Magn Reson Med*. 1996; 35:246–251. [PubMed: 8622590]
7. Cunningham CH, Pauly JM, Nayak KS. Saturated double-angle method for rapid B1+ mapping. *Magn Reson Med*. 2006; 55:1326–1333. [PubMed: 16683260]
8. Yarnykh VL. Actual ip-angle imaging in the pulsed steady state: a method for rapid three-dimensional mapping of the transmitted radiofrequency field. *Magn Reson Med*. 2007; 57:192–200. [PubMed: 17191242]
9. Akoka S, Franconi F, Seguin F, Le Pape A. Radiofrequency map of an NMR coil by imaging. *Magn Reson Imaging*. 1993; 11:437–441. [PubMed: 8505878]
10. Nehrke K, Bornert P. DREAM—a novel approach for robust, ultrafast, multislice B(1) mapping. *Magn Reson Med*. 2012; 68:1517–1526. [PubMed: 22252850]
11. Klose U. Mapping of the radio frequency magnetic field with a MR snapshot FLASH technique. *Med Phys*. 1992; 19:1099–1104. [PubMed: 1518473]
12. Chung S, Kim D, Breton E, Axel L. Rapid B1+ mapping using a preconditioning RF pulse with TurboFLASH readout. *Magn Reson Med*. 2010; 64:439–446. [PubMed: 20665788]
13. Eggenschwiler F, Kober T, Magill AW, Gruetter R, Marques JP. SA2RAGE: A new sequence for fast B1+ -mapping. *Magn Reson Med*. 2012; 67:1609–1619. [PubMed: 22135168]
14. Dowell NG, Tofts PS. Fast, accurate, and precise mapping of the RF field in vivo using the 180 degrees signal null. *Magn Reson Med*. 2007; 58:622–630. [PubMed: 17763355]
15. Sacolick LI, Wiesinger F, Hancu I, Vogel MW. B1 mapping by Bloch-Siegert shift. *Magn Reson Med*. 2010; 63:1315–1322. [PubMed: 20432302]
16. Park, JY.; Garwood, M. Proceedings of the 16th Annual Meeting of ISMRM. Toronto, Canada: 2008. B1 mapping using phase information created by frequency-modulated pulses; p. 361
17. Hennel, F.; Kohler, S. Proceedings of the 18th Annual Meeting of ISMRM. Stockholm, Sweden: 2010. Improved phase-based adiabatic B1 mapping; p. 237
18. Morrell GR. A phase-sensitive method of ip angle mapping. *Magn Reson Med*. 2008; 60:889–894. [PubMed: 18816809]
19. Chang YV. Rapid B1 mapping using orthogonal, equal-amplitude radio-frequency pulses. *Magn Reson Med*. 2012; 67:718–723. [PubMed: 21713982]
20. Conolly S, Nishimura D, Macovski A. Sweep-diagram analysis of selective adiabatic pulses. *J Magn Reson*. 1989; 83:549–564.
21. Silver MS, Joseph RI, Hoult DI. Highly selective $\pi/2$ and π pulse generation. *J Magn Reson*. 1984; 59:347–351.
22. Conolly S, Glover G, Nishimura D, Macovski A. A reduced power selective adiabatic spin-echo pulse sequence. *Magn Reson Med*. 1991; 18:28–38. [PubMed: 2062239]
23. Pauly J, Le Roux P, Nishimura D, Macovski A. Parameter relations for the Shinnar-Le Roux selective excitation pulse design algorithm. *IEEE Trans Med Imaging*. 1991; 10:53–65. [PubMed: 18222800]
24. Khalighi MM, Rutt BK, Kerr AB. RF pulse optimization for Bloch-Siegert B1+ mapping. *Magn Reson Med*. 2012; 68:857–862. [PubMed: 22144397]
25. Khalighi, MM.; Sacolick, LI.; Rutt, BK. Proceedings of the 18th Annual Meeting of ISMRM. Stockholm, Sweden: 2010. Signal to noise ratio analysis of Bloch-Siegert B1+ Mapping; p. 2842
26. Haacke, EM.; Brown, RW.; Thompson, MR.; Venkatesan, R. Magnetic resonance imaging: physical principles and sequence design. 1st ed.. New York: Wiley-Liss; 1999. p. 375
27. Garwood M, DelaBarre L. The return of the frequency sweep: designing adiabatic pulses for contemporary NMR. *J Magn Reson*. 2001; 153:155–177. [PubMed: 11740891]

28. Bernstein, MA.; King, KF.; Zhou, XJ. Handbook of MRI pulse sequences. Burlington, MA: Elsevier; 2004. p. 195
29. Tannús A, Garwood M. Improved performance of frequency-swept pulses using offset-independent adiabaticity. J Magn Reson A. 1996; 120:133–137.
30. Nehrke K, Börnert P. Eigenmode analysis of transmit coil array for tailored B1 mapping. Magn Reson Med. 2010; 63:754–764. [PubMed: 20146236]

Figure 1.

The BEAR sequence: two sech pulses generate a twice-refocused spin-echo. The refocusing pulses can be made selective in Z or Y for slice- or volumetric- imaging. An initial 42° excitation pulse was used for all sequences. A second image is acquired with the order of the AFP pulses reversed (shown in dashed lines).

Figure 2.

(a) BEAR's signal phase dependence on $B_{1,\text{nom}}$ and δ determined by Bloch simulations (solid), and analytical solution φ (dashed). The BS phase dependence on B_1 for an 8-ms 4-kHz-offset Fermi pulse is shown for reference (dotted). (b) Efficiency comparison between BEAR and BS (dotted). For (b,e-f) the solid, dashed, dot-dashed lines are for BEAR with $\delta = 0.7, 0.8, 0.9$. (c) Magnitude and (d) phase plots for Bloch simulations of the slice profile for BEAR. The adiabatic threshold is marked at $B_1 = 0.119$ G, with $\delta = 0.8$. (e) BEAR $B_1 S$ values show that $B_1 S > 1$ rad for a large B_1 range. (f) BEAR B_1 NR ratio as a function of B_1 . (g) The percent B_1 difference between the slice profile (d) and the on-resonance phase variation (a) for $\delta = 0.8$, where blue data is outside the window of interest.

Figure 3.

(a–b) Mean values and standard deviations of measured $B_1/B_{1,max}$ values using BEAR (blue) and BS (red) as a function of transmitter attenuation. The grey line plots ideal conditions ($x=y$). Two different regions of interest of 5×5 pixels were used: (a) high B_1 region with $B_{1,max} = 0.2041$ G, (b) low B_1 region with $B_{1,max} = 0.1976$ G. (c) Percent difference between mean measured $B_1/B_{1,max}$ and transmitter attenuation for BEAR (solid) and BS (dotted), where thicker lines represent the high B_1 region. (d) Example phantom B_1

map for BEAR (top) and BS (bottom), with the two regions of interest indicated by black boxes.

Figure 4.

A comparison of B_1 maps for a phantom at 1.5T: (a) BS method, (b,c) BEAR method with a TR of 500 ms (b) and 100 ms (c), and (d) BEAR 500 ms deviation from BS (RMS = 0.0011 G, $B_{1,nom} = 0.20$ G). (b,e,f) BEAR maps using different values for δ : 0.7 (e), 0.8 (b) and 0.9 (f). (b,g,h) A comparison of BEAR B_1 maps for different $B_{1,nom}$ values: 0.20 G (b), 0.15 G (g) and 0.10 G (h), where the maps in (g,h) are scaled for easier comparison to (b).

Figure 5.

A comparison of B_1 *in vivo* maps at 1.5T (top), 3T (middle), 7T (bottom). (a) Magnitude images from the BEAR sequence. (b) BEAR and (c) BS maps, and (d) BEAR deviations from BS. RMS values of 0.0131 G (1.5T), 0.0158 G (3T), and 0.0228 G (7T).

Figure 6.

In vivo B_1 maps at 1.5T and 3T for: (left) slice and (right) volumetric scans. (b,d) B_1 projections (black) and computed projections (gray) of (a,c), with average difference of 0.75% (1.5T) and 1.6% (3T) for slice scans; 1.2% (1.5T) and 0.56% (3T) for volumetric scans.

Table 1

Polynomial fit values of $B_1(\varphi) = a_0 + a_1\varphi + a_2\varphi^2$ for AFP pulses with $T = 12$ ms, $\beta = 800$ rad/s, $\mu = 5.5$ and δ as shown in column 1.

δ	S_{BEAR} (rad/G)	a_0 (G)	a_1 (G/rad)	a_2 (G/rad ²)	Range (G)	Max error (%)
0.7	56.74	0.05446	0.01883	-0.00006	[0.094, 0.4] [0.134, 0.4]	1.581 0.275
		0.05950	0.01762	-	[0.094, 0.4] [0.134, 0.4]	4.425 0.656
0.8	38.06	0.05174	0.02806	-0.00012	[0.095, 0.4] [0.119, 0.4]	0.720 0.165
		0.05690	0.02628	-	[0.095, 0.4] [0.119, 0.4]	3.439 1.111
0.9	19.13	0.04917	0.05593	-0.00049	[0.098, 0.4] [0.109, 0.4]	0.177 0.089
		0.05469	0.05227	-	[0.098, 0.4] [0.109, 0.4]	2.823 1.822