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Adiabatic RF Pulse Design for Bloch-Siegert B_1^+ Mapping

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Abstract

The Bloch-Siegert (B-S) B_1^+ mapping method has been shown to be fast and accurate, yet it suffers from high SAR and moderately long TE. An adiabatic RF pulse design is introduced here for optimizing the off-resonant B-S RF pulse in order to achieve more B-S B_1^+ measurement sensitivity for a given pulse width. The extra sensitivity can be used for higher Angle to Noise Ratio (ANR) B_1^+ maps or traded off for faster scans. Using numerical simulations and phantom experiments, it is shown that a numerically optimized 2ms adiabatic B-S pulse is 2.5 times more efficient than a conventional 6ms Fermi shaped B-S pulse. The adiabatic B-S pulse performance is validated in a phantom and in vivo brain B_1^+ mapping at 3T and 7T are shown.

Keywords

B_1^+ mapping; flip angle; Bloch-Siegert shift; off resonance; parallel transmit; Adiabatic RF pulse design

INTRODUCTION

In magnetic resonance imaging, mapping of the transmitted radio frequency magnetic field, usually designated as B_1^+ , has a number of applications such as coil design validation, transmit gain calibration or parallel transmit pulse design (1, 2). It can also be used in post processing of quantitative tissue properties imaging such as relaxometry (3).

B_1^+ mapping methods can be broadly categorized into two groups: magnitude based methods such as (4–8) and phase based methods such as (9,10). B_1^+ mapping methods can be prone to long scan times, errors due to imperfect spoiling, unaccounted T1, T2 and B_0 dependencies or errors due to mismatched slab profiles between two or more selective excitations used to compute B_1^+ .

It has been shown that the Bloch-Siebert B_1^+ mapping method (10) is an accurate and efficient phase based method, which is largely insensitive to tissue properties such as T1 and T2, and produces high angle-to-noise-ratio (ANR) maps over a wide range of B_1^+ amplitudes (11). One significant advantage of B-S B_1^+ mapping over other methods in a parallel transmit context is that it decouples the B_1^+ encoding phase of the imaging sequence from the excitation (imaging) phase. This allows independent optimization of imaging and B-S B_1^+ encoding characteristics of the sequence. In a parallel transmit context, this allows transmission on all channels simultaneously during the imaging phase of the sequence while only encoding one channel at a time during the B-S encoding phase, which increases ANR especially in regions of low B_1^+ amplitude (11).

In spite of these advantages, the B-S B_1^+ mapping method still suffers from a moderately long echo time and high RF deposition (SAR) which limits the shortest achievable scan time. Both of these limitations have been recently addressed by the design of short optimized B-S pulses (12). These short B-S pulses are placed at a fixed frequency offset and the pulse envelope has been designed to limit on-resonant excitation while maximizing the Bloch-Siebert phase shift to create high ANR maps.

In the present work, we introduce a new approach for designing *adiabatic* B-S pulses. Unlike our previous B-S pulse design method (12), in which the resonance frequency offset was fixed, here we permit the frequency offset to be time-variable and we employ concepts of adiabatic pulse design. The adiabatic B-S pulse is designed to create the maximum B-S phase shift for a given pulse width and constraint on in-band excitation in order to create the highest quality B_1^+ maps. We demonstrate that these adiabatic Bloch-Siebert pulses offer a substantial improvement in B_1^+ mapping efficiency.

THEORY

We define the Bloch-Siebert pulse to include both amplitude and frequency modulation:

$$B_1(t) = A(t)e^{i\Delta\omega_{RF}(t)t} \quad [1]$$

Where $A(t)$ is the amplitude and $\omega_{RF}(t)$ is the time-varying off-resonance frequency of the B-S pulse. Assuming that the B-S RF pulse is adiabatic, and that magnetization is initially perpendicular to the effective RF field, then the accrued phase of the magnetization will be given by (13):

$$\varphi_{BS} = \int_0^T \gamma |B_{eff}(t)| dt - \int_0^T \Delta\omega_{RF}(t) dt \quad [2]$$

$$|B_{eff}(t)| = \sqrt{A^2(t) + (\Delta\omega_{RF}(t)/\gamma)^2} \quad [3]$$

where T is the total pulse width and γ is the gyromagnetic ratio. The first term in equation 2 describes the phase accrual of the spin-locked magnetization in the doubly-rotating frame,

while the second term corrects for the phase shift between the doubly-rotating frame and standard rotating frame of reference. By substituting equation 3 in 2 and calculating ϕ_{BS}/ω_{RF} , we have:

$$\frac{\partial \phi_{BS}}{\partial \Delta \omega_{RF}} = \int_0^T \frac{\Delta \omega_{RF} - \sqrt{\Delta \omega_{RF}^2 + \gamma^2 A(t)^2}}{\sqrt{\Delta \omega_{RF}^2 + \gamma^2 A(t)^2}} dt < 0 \quad [4]$$

Equation 4 shows that ϕ_{BS} is a monotonically descending function with regards to $\omega_{RF}(t)$ so in order to maximize B-S phase shift, the frequency offset of the B-S pulse should be minimized; practically speaking, it should get as close as possible to resonance without causing in-band excitation greater than a predefined tolerance level. An intuitive solution is to use a frequency swept pulse that begins and ends far off resonance but sweeps the frequency adiabatically towards, and then away from resonance. In this way, the in-band magnetization can be kept in spin-lock (14), thus minimizing any undesired excitation while maximizing the B-S phase shift (Fig. 1-c). To satisfy the adiabatic condition, the frequency sweep rate should be less than γB_{eff} or:

$$\eta = \frac{\gamma |B_{eff}|}{|d\psi/dt|} \gg 1 \quad [5]$$

where ψ is the instantaneous angle of the effective field, given by equation 6 and $d\psi/dt$ is the frequency sweep rate indicating how fast the B-S pulse is approaching resonance.

$$\psi = \tan^{-1} \left(\frac{\gamma A(t)}{\Delta \omega_{RF}(t)} \right) \quad [6]$$

We restrict the pulse to be symmetric in time, with the frequency offset approaching resonance and then returning symmetrically back to the original location in reverse. We additionally limit the maximum amplitude of the pulse, which is defined by the system peak B_1^+ limit, B_{1p} ; in other words, we assume that:

$$A(t) \leq B_{1p} \quad \forall t \quad [7]$$

As can be seen from equations 2 and 3, in order to maximize ϕ_{BS} , $A(t)$ has to be maximized, so we set the amplitude of the B-S pulse to a rect pulse with maximum amplitude B_{1p} . As it was shown by equation 4 to maximize ϕ_{BS} , ω_{RF} has to be minimized, which can be achieved by maximizing $d\psi/dt$. The maximum of $d\psi/dt$ is governed by the adiabatic factor given in equation 5. So in order to maximize $d\psi/dt$, we set the adiabatic factor to a constant minimum value, K , where K is a design parameter which influences the maximum in-band perturbation.

$$\eta = \frac{\gamma |B_{eff}|}{|d\psi/dt|} = K \quad [8]$$

Let's assume $u(t) = \gamma A(t) / \omega_{RF}(t)$ and rewrite equations 6, 3 and 8 again:

$$\psi = \tan^{-1}(u) \Rightarrow \frac{d\psi}{dt} = \frac{du/dt}{1+u^2} \quad [9]$$

$$\gamma |B_{eff}| = \Delta\omega_{RF}(t) \sqrt{1 + \left(\frac{\gamma A(t)}{\Delta\omega_{RF}(t)} \right)^2} = \Delta\omega_{RF}(t) \sqrt{1+u^2} \quad [10]$$

$$\eta = \frac{\Delta\omega_{RF}(t) \sqrt{1+u^2}(1+u^2)}{du/dt} = K \Rightarrow \Delta\omega_{RF}(t) = K \frac{du/dt}{(1+u^2)^{3/2}} \quad [11]$$

Since we assumed a rect shape for the pulse envelope, we can write:

$$A(t) = B_{1p} \Rightarrow \Delta\omega_{RF}(t) = \frac{\gamma B_{1p}}{u(t)} \quad [12]$$

Now solving for $u(t)$ in equations 11 and 12, we will have:

$$\frac{\gamma B_{1p}}{u(t)} = K \frac{du/dt}{(1+u^2)^{3/2}} \Rightarrow u(t) = \sqrt{\frac{1}{\left(1 - \frac{\gamma B_{1p}}{K} t\right)^2} - 1} \quad [13]$$

So the B-S pulse can be written as:

$$A(t) = B_{1p} \quad \Delta\omega_{RF}(t) = \frac{\gamma B_{1p}}{\sqrt{\left(1 - \frac{\gamma B_{1p}}{K} t\right)^2 - 1}}, \quad t=0:T/2 \quad [14]$$

We can also write $\omega_{RF}(t)$ in terms of $\psi(t)$:

$$\psi(t) = \tan^{-1}(u) \Rightarrow \psi(t) = \cos^{-1} \left(1 - \frac{\gamma B_{1p}}{K} t \right) \quad [15]$$

$$\psi_0 = \psi(t)|_{t=T/2} = \cos^{-1} \left(1 - \frac{\gamma B_{1p} T}{2K} \right) \quad [16]$$

$$\Delta\omega_{RF}(t) = \begin{cases} \gamma B_{1p} \cot(\psi) & \text{if } 0 < t \leq \frac{T}{2} \\ \gamma B_{1p} \cot(2\psi_0 - \psi) & \text{if } \frac{T}{2} \leq t < T \end{cases} \quad [17]$$

An example of the above-defined analytical adiabatic B-S pulse is shown in Fig. 1-a,b.

The final step in the pulse design is to refine the analytic solution with numerical optimization to improve the suppression of undesired in-band excitation. The method used is similar to that described in (12), where the B-S pulse energy is maximized while constraining the peak B_1^+ and in-band excitation, but differs in that it allows for complex values in the solution. The quadratic programming function from Matlab (The Mathworks, Natick, USA) is used to find a local minima satisfying:

$$\begin{aligned} &\text{Minimize } (-x^*Ix) \text{ such that} \\ &|Wx| < \delta, 0 < x(n) < 2\pi\gamma\Delta t B_{1p} \quad n=1 \dots N \quad [18] \end{aligned}$$

where $x(n)=2\pi\gamma\Delta t B_1^+(n)$ are the N scaled samples of the fixed-duration RF pulse, I is the identity matrix, W is a DFT matrix evaluating the stopband performance of the pulse, δ is the stopband ripple and t is the duration of each RF sample so that total pulse width $T = N t$. We allow for complex $x(n)$ (in other words, frequency modulated) pulses and repeatedly seed the optimization with the analytical ABS pulse shifted by different frequency offsets in the range of 0–1 kHz to get a series of local minima solutions. A Bloch simulation is then performed for each pulse to find the true cut-off frequency and then the pulse is shifted in frequency to satisfy the $[-600 \ 600] \text{ Hz}$ in-band excitation constraint. Finally using Bloch simulation, the B-S phase shift is calculated for each pulse and the pulse that maximizes the B-S phase shift is selected from the family of results. T_1 and T_2 relaxation were not considered during in the Bloch simulations during the design process because of the very short ABS pulse width. Subsequent Bloch simulations including T_1 , T_2 relaxation effects showed that there was negligible dependence of the B-S phase shift on T_1 , T_2 relaxation times (for T_1 , T_2 values greater than the nominal pulse width).

As has been discussed in (10), in order to accurately measure the B-S phase shift in the presence of other sources of image phase, we use a phase differencing technique, in which the B-S pulse is played at $\pm \omega_{RF}$ and the phase difference between the two measurements calculated so that any common mode phase in the images is removed. For the adiabatic B-S pulse, in order to offset the pulse at $-\omega_{RF}$ the ABS pulse can simply be conjugated.

An efficiency metric given by the following equation was introduced in (12) for comparing different B-S pulses:

$$\Gamma = \frac{\Delta\varphi_{BS}}{E_{BS}} e^{-T/T_2^*} \quad [19]$$

where Γ is the B-S pulse efficiency and E_{BS} is the B-S pulse energy. This measure can be used to compare the efficiency of various B-S pulses by looking at the B-S phase shift generated per unit energy of the pulse and takes into account the T_2^* -dependent signal loss which results from the increased echo time due to the insertion of B-S pulse into the sequence.

METHODS

A 2ms adiabatic B-S pulse was designed using equation [14] with the assumption of $K=42$ and $B_{1p}=20\mu T$. K was selected so that the in-band excitation of the adiabatic B-S pulse was less than -20dB over a conservative range of in-band frequencies (typical of what we would specify at 7T) $[-600 + 600]\text{Hz}$. The frequency response of the designed adiabatic B-S pulse was calculated over a range of B-S pulse amplitudes ($[1\ 20]\mu T$) using Bloch simulation. This pulse was further optimized using the numerical optimization strategy described above, and was compared to the analytical ABS pulse in terms of in-band excitation and B-S phase shift. The numerically optimized 2ms ABS pulse was then compared with the previously used and published 6ms Fermi-shaped B-S pulse. Since the ABS pulse frequency offset is much closer to resonance than the 6ms Fermi pulse (10), the subtraction approach used to eliminate first order B_0 dependence is no longer accurate enough, and therefore the Bloch-simulated phase-shift dependence on B_1 and B_0 was tabulated in a 2D look-up table. B_1 maps were calculated by linear interpolation into this table, given the measured B-S phase shift and an acquired B_0 map.

The B-S pulse was integrated into a conventional gradient echo pulse sequence (10). To minimize artifacts due to in-band excitation by the B-S pulse, crushers were added to the slice-select gradient, before and after B-S pulse to generate at least 2π phase shift across the voxel (12). Using this modified gradient echo sequence, the numerically optimized 2ms ABS pulse was compared to the conventional and published 6ms width/4kHz offset Fermi-shaped B-S pulse (10) for B_1^+ mapping on a 7T scanner (GE Healthcare, Waukesha, WI). Phantom studies, using a 1% copper sulfate filled head-neck shaped phantom, were acquired using the following parameters: $\text{TR}=50\text{ms}$, flip angle= 40deg , $\text{FOV}=24\text{cm}$, thickness= 5mm and 64×64 matrix. The minimum TE was 6.3ms for the ABS pulse and 10.3ms for the Fermi pulse. The B_0 map was determined from two images with different echo times ($\text{TE}=1\text{ms}$) with otherwise identical scan parameters. The B_1^+ mapping sequence was repeated 20 times for each pulse, from which datasets mean B_1^+ and angle-to-noise ratio maps were computed with noise defined as the standard deviation computed across the 20 repeats. The ABS pulse was used for whole brain B_1^+ mapping on a volunteer on a 3T scanner (GE Healthcare, Waukesha WI) with the same sequence with the exception of a longer TR ($\text{TR}=450\text{ms}$) and smaller flip angle ($\text{FA}=20\text{deg}$). TR was chosen so that all slices could fit in one interleaved scan. The total scan time for collecting both B_1^+ and B_0 maps was 2 min.

To evaluate the accuracy of the proposed method, the modified gradient echo sequence was used to measure B_1^+ in a 10cm diameter silicon oil ball phantom on a 7T scanner (GE Healthcare, Waukesha, USA), using the following parameters: $\text{TR}=500\text{ms}$, $\text{TE}=6\text{ms}$, flip angle= 90deg , $\text{FOV}=24\text{cm}$, thickness= 5mm and 64×64 matrix. The ABS pulse amplitude was set to 2, 5, 10, 15 and $20\mu T$ using the scanner's transmit gain calibration tool and by controlling the RF amplitude. By changing the center frequency, off-resonance conditions of -600 , -500 , -300 , 0 , 300 , 500 and 600Hz were created throughout the uniform phantom.

The ABS pulse amplitude was measured by the B-S method over the stated B_1^+ and B_0 ranges; results were averaged over the small phantom.

The B-S pulse was also integrated into a spiral sequence (15) and whole brain B_1^+ maps of a volunteer on a 7T scanner (GE Healthcare, Waukesha, WI) were acquired using the numerically optimized 2ms ABS pulse with single shot spiral, 4096 points, effective resolution=4mm, FA=60deg, FOV=24cm, 25 slices, slice thickness=5mm, readout bandwidth=±83.3kHz, TE=5.9ms, with total scan time=4s. The B_0 map was determined from two single shot spiral images with $TE=1ms$, using otherwise identical scan parameters. All scans used a Nova 32ch head coil (Nova Medical, Wilmington, MA) and all volunteers were scanned in accordance with institutional review board guidelines for in vivo research, and provided informed consent.

RESULTS

The analytically designed 2ms ABS pulse with $K=42$ and $B_{1p}=20\mu T$ is shown in Figure 1-a,b. The frequency sweep of this pulse is shown in Figure 1-c. This pulse generates 211 degree of B-S phase shift and has an efficiency of $238 \text{ deg}(\mu T)^{-2}s^{-1}$ assuming $T_2^*=20ms$. The frequency response of this pulse was calculated over a range of B-S pulse amplitudes ($[1 \text{ } 20]\mu T$) using Bloch simulation and is shown in Figure 1-d. Because B_1^+ is strongly non-uniform at high field and in the case of single transmit channel B_1^+ mapping can change from 0 to B_{1p} , the in-band excitation of the B-S pulse was calculated over a range of B_1^+ amplitudes i.e. $[1\mu T \text{ } B_{1p}]$. As can be seen from this figure, the in-band excitation of the designed B-S pulse is about -20dB or 10%. In order to decrease the in-band excitation, one has to increase the adiabaticity of the pulse by increasing K ; this will decrease the B-S phase shift and therefore the final ANR.

The subsequent numerical optimization of the analytic B-S pulse was able to reduce the in-band excitation while maintaining a similar frequency sweep and pulse efficiency. The numerically optimized 2ms ABS pulse is shown in Figure 2-a,b with its frequency sweep shown in Figure 2-c and its frequency response over the same range of B-S pulse amplitudes shown in Figure 2-d. This pulse is similar to the analytical solution, generating 176 degree of B-S phase shift with efficiency of $292 \text{ deg}(\mu T)^{-2}s^{-1}$ assuming $T_2^*=20ms$ and its in-band excitation is less than -40 dB, which will result in less ghosting artifact and therefore better ANR maps compared to the equivalent width analytical design. In comparison, the $20\mu T$ 6ms Fermi shaped B-S pulse and the $20\mu T$ 2ms magnitude-optimized B-S pulse at a fixed 2850Hz frequency offset, as described in (12), generate 251 and 114 degree of B-S phase with the efficiency of 119 and 205 $\text{deg}(\mu T)^{-2}s^{-1}$ respectively, assuming $T_2^*=20ms$. The numerically optimized 2ms ABS pulse is 150% and 42% more efficient than these two pulses, respectively.

A 4ms ABS pulse was designed by our numerical method and compared to the 4ms Quad pulse introduced in (12). The 4ms ABS pulse with $20\mu T$ peak amplitude generates 494 degree B-S phase shift with an efficiency of $311 \text{ deg}(\mu T)^{-2}s^{-1}$ assuming $T_2^*=20ms$, compared to 329 degree B-S phase shift with an efficiency of $264 \text{ deg}(\mu T)^{-2}s^{-1}$ for the 4ms

Quad pulse with the same peak amplitude, demonstrating a 50% increase in B-S phase shift and 18% improvement in overall efficiency.

Figure 3 illustrates the calculated B-S phase for the 2ms numerically optimized ABS pulse, over a range of B_1^+ of $[0\ 30]\ \mu T$ and a range of B_0 of $[-1\ 1]\ kHz$. We used 101 samples in B_1^+ and 101 samples in B_0 to represent the phase dependence.

Table 1 shows the measurement error of the proposed method with the 2ms numerically optimized ABS pulse over the range of different B_1^+ and B_0 values using a small silicon oil phantom. The mean B_1^+ value shown in this table was obtained by averaging over the phantom; the uniformity across the phantom was better than 2% for all experiments. The measurement error was less than 2.5% across the nominal range of B_1^+ $[2\ 20]\ \mu T$ and B_0 $[-500\ 500]\ Hz$. The error increased to 6% for $B_0 = \pm 600\ Hz$ due to the fact that part of the resonant frequency band of interest now falls outside of the $[-600\ 600]\ Hz$ range used in the ABS pulse design process to constrain the in-band excitation.

Mean B_1^+ and ANR maps produced by the numerically optimized 2ms ABS pulse and conventional 6ms Fermi B-S pulse acquired from the head/neck shaped phantom are shown in Figure 4a. The difference between the B_1^+ maps produced by these two pulses, shown in Figure 4b, is less than 10% everywhere and in the areas with higher SNR is less than 2%. Comparing ANR maps produced with the 2ms ABS pulse and conventional 6ms Fermi B-S pulse, shown in Figure 4c, shows that the 2ms ABS pulse generates higher ANR B_1^+ maps compared to the conventional 6ms Fermi B-S pulse even though it has 2.5 times less energy. The whole brain B_1^+ maps of a volunteer on a 3T scanner using the 2ms ABS pulse in a modified gradient echo sequence is shown in Figure 5, showing the high quality B_1^+ maps that are acquired with minimum TE and minimum TR of the gradient echo sequence within the FDA SAR limits. Whole brain B_1^+ maps using the 2ms ABS pulse in a modified single-shot spiral sequence on a 7T scanner are shown in Figure 6. These results demonstrate that high quality B_1^+ maps can be acquired using single-shot spiral read-out in a very short scan time over the whole brain. The whole brain B_1^+ mapping can be accomplished in very short time frames - as short as 2s/ch - in a multi-channel transmit system using a single-shot spiral sequence modified to incorporate an ABS pulse.

DISCUSSION

To address the increased echo time and high SAR problems of Bloch-Siegert B_1^+ mapping method, we have presented in this paper a new frequency-swept RF pulse which approaches resonance as closely as possible while maintaining minimal in-band excitation. In so doing, we maximize B-S shift and therefore ANR for a fixed pulse width. By maximizing ANR for a fixed pulse width, shorter pulses can be used to achieve high quality B_1^+ maps.

We have presented a closed form expression for the adiabatic Bloch-Siegert pulse assuming constant adiabaticity factor, K , during the RF pulse. Decreasing K will bring the frequency

offset closer to resonance and thereby increasing the B-S phase shift; however, this will increase the unwanted signal coming from in-band excitation and will require more crushing around the Bloch-Siegert pulse in order to prevent image artifact. It will also return some of the transverse magnetization back to the longitudinal axis, which will decrease the signal.

The analytical ABS pulse, which is based on fulfilling the adiabatic condition, does not explicitly constrain the in-band excitation. Our numerical optimization approach for the refinement of the analytical ABS pulse was successful in improving suppression of in-band excitation while maintaining a similar frequency sweep and B-S phase shift to the analytic ABS pulse. However, as our current implementation does not explicitly include the B-S phase shift in the objective function it is possible that optimization could return a solution with substantially degraded B-S phase shift. Future work will include the B-S phase shift explicitly in the objective function as in (16) to potentially make the method more robust. The 2ms optimized pulse described in (16) has a similar performance compared to the 2ms numerically optimized ABS pulse described here in regards to B-S phase shift; however, the ABS pulse deposits 32% less energy and therefore is 32% more efficient.

As its been stated in (10), magnetization transfer (MT) can lead to lower signal levels than would be present in the absence of the B-S pulse. The smaller off-resonant shift of the ABS pulse will act to increase the MT effect, while the reduced power will act to decrease the effect. Experimentally, we found very little change in signal magnitude for images acquired with and without application of the ABS pulse.

We showed that the 2ms ABS pulse generates higher ANR B_1^+ maps in a head-neck phantom compared to the conventional 6ms width/4kHz offset Fermi B-S pulse while it has 2.5 times less energy or SAR and reduces echo time by 4ms. By implementing the new ABS pulse in both conventional gradient echo and spiral sequences and testing it on two high field scanners (3T and 7T), we showed that these sequences are no longer limited by SAR and can be operated with minimum TE and minimum TR for whole brain B_1^+ mapping. Due to the low spatial frequency content in B_1^+ maps, a single-shot spiral sequence is adequate for generating fast high quality B_1^+ maps even at 7T.

CONCLUSIONS

We have introduced the adiabatic Bloch-Siegert pulse design algorithm; this algorithm results in a frequency swept B-S pulse whose frequency approaches resonance, thereby maximizing the B-S phase shift for a fixed pulse width. The time-symmetric adiabatic pulse restores the longitudinal magnetization back to the longitudinal axis to minimize any artifact caused by unwanted in-band excitation, meanwhile the transverse magnetization is spin locked due to the adiabatic property of the B-S pulse and remains unperturbed apart from the B-S phase shift. The new design allows for shorter B-S pulses with the same or better ANR maps compared to the traditional Fermi-shaped B-S pulse; for this reason, the echo time and SAR are reduced substantially and this leads to considerable decrease in scan times.

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Figure 1.

2ms Adiabatic B-S pulse with $K=42$ and $B_{1p}=20\mu T$. (a) B-S pulse amplitude $A(t)$, (b) B-S pulse frequency ω_{RF} , (c) the frequency sweep and (d) the frequency response of the B-S pulse for amplitudes in $[1\ 20]\mu T$. The in-band excitation is less than -20dB over 1200Hz bandwidth. This pulse creates 211 degree of B-S phase shift with the efficiency of $238\text{ deg}(\mu T)^{-2}s^{-1}$ for $T_2^*=20\text{ms}$ and it gets as close as 1476Hz to the resonance frequency.

Figure 2.

2ms Adiabatic B-S pulse designed with MATLAB quadratic programming with $B_{1p}=20\mu T$. (a) B-S pulse amplitude $A(t)$, (b) B-S pulse frequency ω_{RF} , (c) the frequency sweep and (d) the frequency response of the B-S pulse for amplitudes in $[1\ 20]\mu T$. The in-band excitation is less than -40dB over 1200Hz bandwidth. This pulse creates 176 degree of B-S phase shift with the efficiency of $292\text{ deg}(\mu T)^{-2}s^{-1}$ for $T_2^*=20\text{ms}$ and it gets as close as 1438Hz to the resonance frequency.

Figure 3.

Calculated B-S phase for the 2ms numerically optimized ABS pulse, over a range of B_1^+ of $[0\ 30]\ \mu T$ and a range of B_0 of $[-1\ 1]\ kHz$.

Figure 4. Comparing 2ms ABS pulse with conventional 6ms Fermi B-S pulse using a head-neck phantom on a 7T scanner: (a) Mean B_1^+ maps (b) difference between mean B_1^+ maps and (c) ANR of B_1^+ maps.

Figure 5.

Whole brain B_1^+ mapping using 2ms ABS pulse on a 3T scanner with a modified gradient echo sequence, TE=6.3ms, TR=450ms, flip angle=20deg, 64×64 matrix.

Figure 6.

Whole brain B_1^+ mapping using 2ms ABS pulse on a 7T scanner with a modified single-shot spiral sequence, 4096 points, effective resolution=4mm, TE=5.9ms, flip angle=60deg, bandwidth= ± 83.3 kHz and scan time=4s.

Table 1

B-S method accuracy with ABS pulse

B_0 (Hz)	$B_1^+ (\mu T)$							
	2	5	10	15	20			
600	4.3	3.9	5.2	5.1	5.2			
500	0.5	0.2	0.4	2.2	2.5			
300	0.9	0.3	0.2	2.2	2.5			
0	0.0	0.2	0.5	1.9	2.3			
-300	0.0	0.3	0.3	1.7	2.2			
-500	0.0	0.5	0.8	2.2	2.5			
-600	5.2	4.4	4.7	6.1	6.0			

Error(%) between measured $|B_1^+|$ by B-S method using the 2ms numerically designed ABS pulse, and $|B_1^+|$ set by system's gain calibration tool in a small silicon oil phantom at 7T.