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Steady State Double Angle Method for Rapid B_1 Mapping

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Abstract

Purpose: To introduce an accurate, rapid, and practical method for active B_1 field mapping based on the double angle method (DAM) in the steady state (SS) signal regime.

Methods: We introduced and evaluated the performance of the SS-DAM approach to map the B_1 field, and compared the results to those calculated from the conventional DAM approach. Similar to DAM, SS-DAM uses the signal intensity ratio of two magnitude images acquired with different flip angles using the spoiled gradient recalled echo (SPGR) sequence. However, unlike DAM, in SS-DAM, these two SPGR images are acquired with very short repetition time; this allows substantially reduced acquisition time. Numerical, phantom, and *in-vivo* brain imaging analyses, representing a wide range of longitudinal relaxation times and large B_1 variation, were conducted. Methods for further accelerating acquisition were also investigated.

Results: Our results demonstrate the potential of the SS-DAM approach to be applied widely in the clinical setting. B_1 maps derived from SS-DAM were demonstrated to be quantitatively comparable to those derived from DAM, but were derived much more rapidly. Large volume B_1 maps were obtained at a field strength of 3T within clinically acceptable acquisition times.

Conclusions: SS-DAM permits accurate B_1 mapping in the clinical setting, with whole-brain coverage in less than 1 min.

Keywords

Double angle method; Steady state sequence; B_1 mapping; SPGR

1. INTRODUCTION

Quantitative determination of magnetic resonance (MR) parameters is affected by inaccuracies in the radio-frequency (RF) flip angle (FA) (1–9), so that accurate parameter estimation requires calibration of the RF transmit field, B_1 . This has become particularly problematic with the growing use of high-field MRI systems, since at high field, the spatial inhomogeneity of B_1 is increased due to the smaller RF wavelength compared to sample size (10, 11). While several effective B_1 mapping methods based on phase or magnitude imaging have been introduced (12–23), these methods often require complex modeling, long acquisition time, or specialized MRI sequences. Therefore, these methods have not been

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integrated into clinical use, resulting in an ongoing need for a practical, accurate, and fast B_1 mapping method.

The double angle method (DAM) (18) is among the simplest of the available B_1 mapping methods. DAM makes use of the ratio of the signal intensity of two magnitude images acquired respectively with excitation FAs of α and 2α . To mitigate the effect of finite longitudinal relaxation time, T_1 , these images must be acquired with a long repetition time (TR), resulting in a lengthy acquisition. Several variants of DAM and related methods have been proposed to accelerate acquisition, including the saturated DAM (19), DAM with selective pulses (12), the actual flip angle approach (20), the stimulated-echo acquisition mode (21), and the dual refocusing echo acquisition mode (24). However, these methods, while rapid and accurate, require specialized MRI sequences or complex modeling, complicating their integration into clinical investigations.

Inspired by the work of Dowell and Tofts (15), we introduce a new variant of DAM, SS-DAM, that operates in the steady state (SS) regime, that is, using a very short TR. SS-DAM exploits the linearity of the spoiled gradient recalled echo (SPGR) signal amplitude as a function of FA, over a wide range of FAs around 180° . We show that in this regime, the ratio of the intensities of two images acquired at different FAs, as described below, is independent of T_1 for a wide range of B_1 values, and that this permits a wide range of B_1 variation to be mapped with simple modeling. We evaluate the performance of SS-DAM using numerical simulations, and phantom and *in-vivo* experiments at 3T. B_1 maps derived from SS-DAM were compared to those derived from the conventional DAM approach. We also evaluate and discuss strategies for further acceleration of B_1 mapping.

2. THEORY

Under perfect spoiling of transverse magnetization, the magnitude of the SPGR signal is conventionally given by:

$$S_{\text{SPGR}}(\zeta\alpha, \text{TR}) = \left| S_0 \exp(-\text{TE}/T_2^*) \frac{1 - \exp(-\text{TR}/T_1)}{1 - \exp(-\text{TR}/T_1) \cos(\zeta\alpha)} \sin(\zeta\alpha) \right|, \quad [1]$$

where the dimensionless quantity ζ denotes the local, voxel-wise, B_1 scaling factor, S_0 represents the signal amplitude incorporating effects of proton density, diffusion, and machine factors, TE is the echo time, T_2^* is the apparent transverse relaxation time, α is the nominal excitation FA, TR is the repetition time, and T_1 is the longitudinal relaxation time.

Figure 1 shows that S_{SPGR} varies linearly as a function of α over a large range of α centered at 180° , with a slope depending on TR/T_1 . In fact, in this linear regime, it is clear that the effect of T_1 is only to change the slope of this relationship, with the y-intercept remaining at zero for $\alpha = 180^\circ$. Therefore, the intensity ratio, Q , of two images acquired with different values of α within this linear region is approximately independent of T_1 , and depends only on B_1 .

To calculate ζ voxel-wise, $Q = S_{\text{SPGR}}(\zeta(180^\circ - \theta_1), \text{TR}) / S_{\text{SPGR}}(\zeta(180^\circ - \theta_2), \text{TR})$ can first be computed for a large range of ζ , assuming any value of T_1 (e.g. $T_1 = 2000$ ms) and for specified values of θ_1 and θ_2 , in order to create a look-up table of Q vs. ζ . The experimental value of Q , Q_{exp} , is determined voxel-wise from SPGR images as above. ζ is then estimated in each voxel using a linear interpolation based on the pre-defined look-up table. In this study we assumed for simplicity that $\theta_1 = \theta_2 = \theta$. Note that the values of Q are undefined at $\zeta = 180^\circ / (180^\circ + \theta)$ and $\zeta = 360^\circ / (180^\circ + \theta)$ defining the limits of the measurable ζ . Of note, we used the MATLAB function “*interp1*” to perform the linear interpolation. If the experimental value of Q is outside the range of the simulated Q values, due to large variations in ζ or low signal-to-noise ratio (SNR) of the SPGR images, ζ cannot be estimated. All numerical calculations were performed with MATLAB (MathWorks, Natick, MA, USA).

3. METHODS

3.1. Signal ratio Q as a function of ζ for different θ and T_1 values

To verify its independence of T_1 and demonstrate the effect of θ , Q was computed for TR/T_1 values of 0.2, 0.1, 0.05, 0.025, 0.01 and 0.005 and θ values of 40° , 60° , 80° , 100° , and 120° . For each T_1 and θ combination, Q was computed for a large range of ζ , with $0.4 \leq \zeta \leq 1.6$. TR was fixed to 35 ms for all analyses.

We also analyzed the dependence of ζ on the T_1 value used to generate the look-up table of Q vs. ζ . For this, 2D noise-free SPGR T_1 -weighted brain images were generated, using Eq. 1, at FAs of 120° and 240° (i.e. $180^\circ \pm 60^\circ$), $\text{TR} = 35$ ms, and using pre-generated input 2D T_1 and ζ maps. The input T_1 map was generated from synthetic structural images obtained from BrainWeb (25) with values of 800 ms, 1500 ms, and 4000 ms corresponding, respectively, to white matter (WM), gray matter (GM), and cerebrospinal fluid (CSF). The input ζ map was generated assuming a 2D radially symmetric Gaussian distribution, with central value of 1.4 and decreasing by 1.5 standard deviation (SD) to a value of 0.7 at the mid-point of the sides of the imaging matrix, encompassing typical B_1 variation at 1.5T and 3T. S_0 was set to unity and TE to 0 ms. Four different look-up tables were generated with T_1 values of 500 ms, 1000 ms, 2000 ms, and 4000 ms. ζ maps were then estimated using each of these look-up tables. Relative error (RE) maps were calculated voxel-wise according to $\text{RE}(\%) = 100 \times |\zeta_{\text{est}} - \zeta_{\text{ref}}| / \zeta_{\text{ref}}$, where ζ_{est} and ζ_{ref} are the estimated and reference (input) ζ maps at a given voxel, respectively. Mean and SD RE values calculated over all brain voxels were also computed.

3.2. Effect of imperfect transverse magnetization spoiling on Q

The SPGR signal given by Eq. 1 assumes complete spoiling of the transverse magnetization before each RF pulse. Spoiling is most usually performed through use of crusher gradients applied before each excitation RF pulse. However, this method may be insufficient to fully dephase transverse magnetization, so that the use of crusher gradients combined with phase incrementation of RF excitation pulses has been proposed to achieve higher spoiling efficiency (26–28). In fact, it has been demonstrated that progressively incrementing the phase of the applied RF improves the performance of spoiling and therefore achieving an SS

SPGR signal accurately modeled by Eq. 1 (26–28). Using the Bloch equations and the isochromat summation technique, we analyzed the potential effect of these spoiling methods on Q .

Briefly, the magnetization vector for an isochromat of phase ψ_c with magnetization $\mathbf{M}_{i+1}^-(\psi_c)$ before (sign -) the $(i+1)$ -th RF pulse is given by (27, 29–31):

$$\mathbf{M}_{i+1}^-(\psi_c) = \mathbf{E}(\psi_c)\mathbf{M}_i^+(\psi_c) + \mathbf{E}_0, \quad [2]$$

where $\mathbf{M}_i^+(\psi_c) = \mathbf{R}_{\alpha, \varphi_i} \mathbf{M}_i^-(\psi_c)$ represents the magnetization vector after (sign +) the i -th RF pulse, $\mathbf{E}_0 = (1 - E_1)\mathbf{M}_0 = (0, 0, 1 - E_1)$, where $\mathbf{M}_0$ is the equilibrium magnetization vector,

$$\text{and } \mathbf{E}(\psi_c) = \text{diag}(E_2, E_2, E_1) \mathbf{R}_z(\psi_c) = \begin{pmatrix} E_2 \cos \psi_c & E_2 \sin \psi_c & 0 \\ -E_2 \sin \psi_c & E_2 \cos \psi_c & 0 \\ 0 & 0 & E_1 \end{pmatrix} \text{ with } E_1 = \exp(-\text{TR}/T_1) \text{ and}$$

$E_2 = \exp(-\text{TR}/T_2^*)$. The effect of the crusher gradients, applied before each RF pulse, is to uniformly distribute the isochromat phases in each voxel between 0 and 2π , so that $\psi_c = 2\pi c/N_c$ ($c = 1, 2, \dots, N_c$), where N_c is the total number of isochromats in this voxel. $\mathbf{R}_{\alpha, \varphi_i}$ is the rotation matrix resulting from the application of an RF pulse of angle α and phase φ_i given by:

$$\mathbf{R}_{\alpha, \varphi_i} = \begin{pmatrix} \cos(\zeta\alpha)\sin^2\varphi_i + \cos^2\varphi_i & (1 - \cos(\zeta\alpha))\sin\varphi_i\cos\varphi_i & -\sin(\zeta\alpha)\sin\varphi_i \\ (1 - \cos(\zeta\alpha))\sin\varphi_i\cos\varphi_i & \cos(\zeta\alpha)\cos^2\varphi_i + \sin^2\varphi_i & \sin(\zeta\alpha)\cos\varphi_i \\ \sin(\zeta\alpha)\sin\varphi_i & -\sin(\zeta\alpha)\cos\varphi_i & \cos(\zeta\alpha) \end{pmatrix}$$

Finally, the steady state signal, $S_{SS,i}^+$, after the i -th RF pulse is given by the magnitude of the net magnetization

$$S_{SS,i}^+ = \sqrt{\left(\frac{1}{N_c} \sum_{c=1}^{N_c} M_{x,i}^+(\psi_c) \right)^2 + \left(\frac{1}{N_c} \sum_{c=1}^{N_c} M_{y,i}^+(\psi_c) \right)^2} \quad [3]$$

where M_x and M_y are the transverse components of the magnetization vector $\mathbf{M}$.

Using Eq. 3, we first investigated the number of RF pulses required to reach a steady state using crusher gradients with RF pulses of the same phase (*i.e.* $\varphi_i = 0^\circ$) or crusher gradients combined with RF phase incrementation (*i.e.* $\varphi_i = i(i+1)\varphi$, where φ is the constant phase increment). For each isochromat, Eqs. 2–3 were repeatedly solved until the total magnetization (Eq. 3) reached a SS. The second analysis consisted of calculating the ratio Q for the large ζ variation defined above, from ideal signals (Eq. 1), from SS signals obtained with crusher gradients and constant-phase pulses, or from SS signals obtained with crusher

gradients combined with RF phase incrementation. The third simulation analysis consisted of evaluating the impact of each spoiling method on derived ζ values. T_I -weighted images were generated from SS signals obtained with crusher gradients and constant-phase pulses or from SS signals obtained with crusher gradients combined with RF phase incrementation using the ζ and T_I input maps described above. For each spoiling method, ζ maps, RE maps defined with respect to the ideal, fully-spoiled, images, and corresponding mean and SD RE values calculated over all brain voxels were displayed. All analyses were performed for three different tissue types, namely, WM, GM, and CSF, with representative relaxation times, T_2^*/T_1 , of 50/800 ms, 80/1500 ms, and 150/4000 ms, respectively. Other simulation parameters were $N_c = 360$, $TR = 35$ ms, $\alpha = 180^\circ \pm \theta$ with $\theta = 60^\circ$, and $\phi = 42^\circ$ (27, 32).

3.3. Effect of noise on derived ζ maps using SS-DAM and DAM

This analysis consisted of evaluating the sensitivity of the conventional DAM and of SS-DAM to noise. Zero-mean Gaussian noise of SD σ was added to the simulated T_I -weighted images described above. We analyzed two different implementations of DAM. The first corresponds to the typical use of DAM, with images generated at FAs of 60° and 120° (*i.e.* $\alpha = 60^\circ$). The second uses the flip angles implemented in SS-DAM, that is, 120° and 240° (*i.e.* $\alpha = 120^\circ$). DAM SPGR images were generated using Eq. 1. All other experimental parameters were identical to those used to generate the SS-DAM SPGR images, described above, but with TR fixed at 8000 ms. Defining SNR as the magnitude of the WM signal divided by σ , we generated images with low (SNR = 10), clinical (SNR = 100), and high (SNR = 1000) SNR values. For each of these, RE maps defined with respect to ζ_{ref} and corresponding mean and SD RE values calculated over all brain voxels were displayed.

3.4. Phantom experiments

We evaluated the performance of SS-DAM on a large (9 cm diameter $\times$ 22 cm high) bottle of water doped with copper sulfate to achieve a T_I value of ~ 500 ms. Two sagittal 3D SPGR images were acquired, with FAs of 120° or 240° (*i.e.* $180^\circ \pm 60^\circ$). Acquisition parameters included a SENSE factor of 2 in the left/right direction, $TR = 35$ ms, $TE = 5.3$ ms, field-of-view (FoV) = $220 \text{ mm} \times 220 \text{ mm} \times 180 \text{ mm}$, and acquisition voxel size = $1.8 \text{ mm} \times 1.8 \text{ mm} \times 4 \text{ mm}$ reconstructed to $1.6 \text{ mm} \times 1.6 \text{ mm} \times 1.6 \text{ mm}$. The total acquisition time (TAT) was ~ 3 min. For comparison, ζ maps derived using SS-DAM were compared to map derived using the DAM. The DAM protocol consisted of acquiring two 3D SPGR images with FAs of 120° and 240° , with $TR = 8000$ ms, $TE = 8.1$ ms, FoV = $220 \text{ mm} \times 220 \text{ mm} \times 180 \text{ mm}$, EPI factor = 13, a SENSE factor of 2 in the left/right direction, and acquisition voxel size = $2 \text{ mm} \times 2 \text{ mm} \times 4 \text{ mm}$ reconstructed to $1.6 \text{ mm} \times 1.6 \text{ mm} \times 1.6 \text{ mm}$. TAT was ~ 63 min. For all scans, the slab-select gradient was not applied during the RF pulse to avoid loss of signal intensity toward the edges of the prescribed volume. All experiments were performed on a 3T whole body Philips MRI system (Achieva, Best, The Netherlands) using the internal quadrature body coil for transmission and an eight-channel phased-array head coil for reception. In addition to visual inspection, RE maps were calculated according to $RE(\%) = 100 \times |\zeta_{SS-DAM} - \zeta_{DAM}| / \zeta_{DAM}$, where ζ_{SS-DAM} and ζ_{DAM} are the SS-DAM and DAM ζ maps, respectively. For each slice, corresponding mean and SD RE values were calculated over all phantom regions.

We also evaluated the sensitivity of the SS-DAM and DAM to off-resonance effects. Off-resonance effects manifest particularly in regions with strong susceptibility gradients, including brain regions surrounding the sinuses; this leads to decreased effectiveness of the applied RF pulse (13). To evaluate this effect, we followed the same methodology as described by Dowell and Tofts in their seminal paper (15). Specifically, images were obtained on a large cylindrical phantom (Philips; 9 cm diameter $\times$ 22 cm high) filled with mineral oil (1-decene homopolymer hydrogenated) (15). Different frequency offsets, ν , were used with the transmitter frequency on resonance, that is, $\nu = 0$ Hz, or off resonance at different ν values of ± 100 Hz, ± 200 Hz, and ± 500 Hz. Acquisition parameters were identical to those described above. For all experiments, we used the default Philips sine-Gaussian RF excitation pulse with bandwidth (BW) of 1 KHz and duration of 9 ms. Off-resonance effects were explored, for each $\nu \neq 0$ and each method, by measuring the mean deviation of derived ζ values from images obtained with $\nu \neq 0$ from the value derived from images obtained with $\nu = 0$ Hz in a large region-of-interest (ROI) located at the center of the coil (2 cm $\times$ 2 cm $\times$ 2 cm). Deviation was defined as

$(100/N) \times \sum_{i=1}^N |\zeta_{\Delta\nu=0,i} - \zeta_{\Delta\nu \neq 0,i}| / \zeta_{\Delta\nu=0,i}$, where N is the total number of voxels in the ROI (15).

3.5. In-vivo analysis

To evaluate the SS-DAM *in-vivo*, two sagittal 3D SPGR images were acquired from the brain of two healthy subjects (male, age 52, and female, age 72) at FAs of 120° or 240° (*i.e.* 180° $\pm$ 60°). Parameters included a SENSE factor of 2 in the left/right direction, TR = 35 ms, TE = 5.1 ms, FoV = 260 mm $\times$ 260 mm $\times$ 215 mm, and acquisition voxel size = 2 mm $\times$ 2 mm $\times$ 5 mm reconstructed to 2 mm $\times$ 2 mm $\times$ 2 mm. The TAT was 3 min, leading to ζ mapping with 4.2 seconds/slice. For one participant, an extra image was acquired with FA = 0° to estimate SNR. The approximate SNR, defined as the mean signal value calculated over a large WM region from the image acquired at FA = 120° divided by the signal SD calculated over the same region from the image acquired at 0°, was ~ 290 . Again, ζ maps obtained using SS-DAM were compared to maps derived using DAM. The DAM protocol consisted of acquiring two 3D SPGR images with FAs of 120° and 240°, TR = 8000 ms, TE = 6.2 ms, SENSE factor of 2 in the left/right direction, FoV = 260 mm $\times$ 260 mm $\times$ 215 mm, EPI factor = 11, and acquisition voxel size = 4 mm $\times$ 4 mm $\times$ 5 mm reconstructed to 2 mm $\times$ 2 mm $\times$ 2 mm. The TAT was ~ 45 min. Manual segmentation and mask creation were performed to eliminate non-parenchymal regions. As in the case of the phantom experiments, for all scans on these subjects, the slab-select gradient was not applied during the RF pulse to avoid loss of signal intensity toward the edges of the prescribed volume. Frequency-encoding was applied along the head/foot direction to avoid aliasing from the shoulders and upper body regions. In addition to visual inspection, RE maps were calculated according to $RE(\%) = 100 \times |\zeta_{SS-DAM} - \zeta_{DAM}| / \zeta_{DAM}$, where ζ_{SS-DAM} and ζ_{DAM} are the SS-DAM and DAM ζ maps, respectively. For each slice, corresponding mean and SD RE values were calculated over all brain regions.

Moreover, we investigated three additional imaging acceleration methods in conjunction with SS-DAM. The first method consisted of using half-scan, leading to ζ mapping with 2.9 seconds/slice. The second consisted of using a SENSE factor of 2 in both left/right and

anterior/posterior directions, leading to ζ mapping with 2.4 seconds/slice. The third method consisted of combining both of these, leading to ζ mapping with 1.6 seconds/slice. All other parameters were identical to those described previously. In addition to visual inspection, RE maps were calculated according to $RE(\%) = 100 \times |\zeta_{Accel} - \zeta_{Unaccel}| / \zeta_{Unaccel}$ where ζ_{Accel} and $\zeta_{Unaccel}$ are the ζ maps obtained from accelerated and unaccelerated imaging protocols, respectively. For each slice, corresponding mean and SD RE values were calculated over all brain regions. Here as well, we used the default Philips sine-Gaussian RF excitation pulse with BW of 1 KHz and duration of 9 ms.

For all *in-vivo* acquisitions, the nature of the experimental procedures was fully explained to the subjects, with written informed consent obtained prior to participation. All examinations were performed in compliance with the standards established by our local Institutional Review Board.

4. RESULTS

4.1. Numerical analysis

Figure 2 shows Q as a function of ζ for different TR/T_1 and θ values. As is readily seen, for each θ value, there is a large range of ζ , indicated by the vertical black dashed lines, over which Q is nearly independent of TR/T_1 , and is slowly varying and monotonic as a function of ζ . This permits interpolation of measured values of Q to obtain unambiguous value for ζ . For θ of 40°, 60°, 80°, 100°, and 120°, this range spans 0.8 $\leq \zeta \leq$ 1.3, 0.75 $\leq \zeta \leq$ 1.45, 0.7 $\leq \zeta \leq$ 1.25, 0.625 $\leq \zeta \leq$ 1.15, and 0.6 $\leq \zeta \leq$ 1.1, respectively, corresponding to typical ζ variation in clinical MR imaging (3T). Of note, for θ values larger than ~80°, Q becomes progressively more dependent on TR/T_1 ; this is due to the increased nonlinearity of the SPGR signal (Fig. 1). In the following, we used θ of 60° as this choice provides a large usable range of measurable ζ .

Figure 3 shows simulation results for ζ maps derived using SS-DAM with four different look-up tables of Q vs. ζ , generated with T_1 values of 500 ms, 1000 ms, 2000 ms and 4000 ms. In all cases, ζ maps were very similar to the reference ζ map. This was confirmed quantitatively from the corresponding RE maps, defined with respect to ζ_{ref} showing mean RE values of less than 0.2 %. These results indicate the near-independence of SS-DAM on T_1 . Of note, the red arrows indicate regions with slightly larger RE. We attribute these to local ζ values (*i.e.* < 0.75) outside the ideal range of ζ for $\theta = 60^\circ$ (Fig. 2).

Figure 4 shows simulation results for the number of RF pulses required to reach the SS, and the Q ratio as a function of ζ for different spoiling methods and for the three different tissue types. In all cases, the signal stabilizes and reaches the SS after no more than ~50 RF pulse excitations. This indicates that no dummy scans are required before the 3D SPGR acquisition. Furthermore, as expected, the use of the RF phase incrementation combined with crusher gradients results in better magnetization spoiling as compared to the use of crusher gradients only (26, 27). Most importantly, imperfect spoiling has only a marginal effect on Q ; this may be of particular importance for the widespread applicability of the SS-DAM approach in clinical applications with potentially less well-optimized pulse parameters. Indeed, figure 5 shows that derived ζ maps from SPGR images generated with

the two spoiling methods evaluated, as described above, look virtually identical to the ζ_{ref} map, and exhibit a maximal RE, defined with respect to perfect spoiling, value less than $\sim 1.5\%$ over all brain regions. In addition, we note a slight superiority of the spoiling method using crusher gradients only.

Figure 6 shows simulation results for estimated ζ maps using SS-DAM or DAM from SPGR brain images generated for three different SNR values of 10, 100, and 1000 corresponding to low, clinical, and high SNR, respectively. For all SNRs, estimated ζ maps using the SS-DAM were similar to the ζ_{ref} map. Moreover, quantitative analysis shows that the RE is negligible even at low SNR regime. We note that the higher RE seen in CSF regions at low SNR (*i.e.* SNR = 10) is attributable to the very low signal intensities in the T_I -weighted images in those regions due to the long T_I of CSF. However, even in these extreme conditions, RE remains negligible. These results indicate the robustness of SS-DAM to noise. Here again, the red arrows indicate regions with relatively large RE values. This is due to the fact that the ζ values in those regions were outside the usable range of ζ for $\theta = 60^\circ$ (Fig. 2). Likewise, estimated ζ maps using the DAM were, overall, similar to the reference ζ map except in CSF regions. As expected, in these regions, RE values were large due to the known T_I dependence of the DAM with the relatively short TR used here. Finally, we note slightly better overall results with DAM incorporating $\alpha = 120^\circ$ as compared to the conventional use of DAM with $\alpha = 60^\circ$, as indicated by the mean and SD RE values.

4.2. Phantom analysis

Figure 7 shows SS-DAM and DAM SPGR images and corresponding derived ζ maps obtained from imaging data obtained on the uniform phantom. The SPGR images show large spatial variations in signal intensity across the phantom, as reflected in the ζ maps which exhibit a smooth decrease in ζ as a function of distance from the center of the coil. More importantly, visual inspection, RE maps, defined with respect to the DAM method, and mean and SD RE values show that ζ maps derived using SS-DAM or DAM were very similar in most regions. Small differences seen especially at the edges, as indicated by the red arrows, are due to susceptibility artifacts caused by the low imaging resolution and EPI encoding in the DAM imaging acquisition. We note a rapid field modulation at the lower and upper regions of the phantom; this is likely due to the strong off-resonance effects at the phantom's boundaries and attendant susceptibility gradients.

Figure 8 shows percentage deviation in ζ derived from the SS-DAM or DAM at different RF frequency offsets, ν . As is readily seen, SS-DAM and DAM exhibit similar sensitivity to off-resonance effects, with increased deviation with increased ν . While minimal in the range of $\nu = \pm 250$ Hz, the deviation in derived ζ increases rapidly outside that range. This agrees with the results obtained by Dowell and Tofts (15).

4.3. In-vivo analysis

Figure 9 shows examples of *in-vivo* SS-DAM and DAM SPGR images and corresponding derived ζ maps obtained from the brains of two participants. Here again, visual inspection of the weighted images shows a large spatial variation in overall signal intensity across the brain. This is quantitatively illustrated by the smooth variation seen in the ζ maps.

Furthermore, and more importantly, visual inspection, RE maps, defined with respect to the DAM-derived values, and mean and SD RE values show that the ζ maps derived using SS-DAM are very similar to those derived from the DAM. We note that the small differences seen in CSF and some peripheral regions are due, respectively, to the known T_1 dependence of the DAM given the relatively short TR used here, in agreement with simulation results (Fig. 6), and susceptibility artifacts resulting from the limited imaging resolution and EPI encoding used in the DAM. Finally, the computation times required to generate a whole brain ζ map using SS-DAM and DAM were ~ 26 s and ~ 0.1 s, using a 2.4-GHz computer.

Figure 10 shows examples of ζ maps derived using SS-DAM from SPGR images acquired with three different imaging acceleration methods. Visual inspection shows that the ζ maps derived from these acceleration methods are very similar to the reference ζ maps derived from SS-DAM data acquired without acceleration and which define the standard for the RE calculation in this context. We note that the small differences seen in some brain regions are likely due to motion artifacts resulting from the lengthy acquisition time required to acquire all scans for this study. Combining SENSE acceleration and half-scan imaging permits whole brain ζ mapping in less than 1 min using SS-DAM.

5. DISCUSSION

The SS-DAM provides a rapid and robust approach to B_1 field mapping. SS-DAM uses the signal intensity ratio of two magnitude SPGR images measured with different excitation FAs and a very short TR and takes advantage of the linearity of the intensity around $FA = 180^\circ$. While the steady-state SPGR signal equation (Eq. 1) is dependent on T_1 , the ratio, Q of two SPGR images acquired at FAs disposed around 180° is virtually independent of T_1 . When scaling errors, ζ , in the FAs are introduced, then Q becomes dependent of T_1 . However, this dependence remains marginal for a reasonable range of ζ (Figs. 1–6). The required SPGR (*i.e.* FLASH, FFE) sequence is available on all preclinical and clinical MRI systems. Our extensive numerical, phantom, and *in-vivo* analyses demonstrated that the accuracy of the B_1 maps derived from SS-DAM compares favorably with those derived from the conventional DAM (Fig. 7).

Our results show that SS-DAM is very robust to noise, indicating that accurate B_1 mapping may be possible with further imaging acceleration. Even without acceleration, we demonstrated that use of half-scan and SENSE reconstruction (33–35) permits whole brain B_1 mapping in less than 1 min (Fig. 10). Other acceleration strategies, including EPI and spiral encodings, might be used. We note that the total acquisition time of SS-DAM could be reduced by decreasing the FoV or shortening TR, although these strategies may run up against specific absorption rate (SAR) limits, given the large FAs used here. In this work, we used the minimum TR allowed by the scanner while maintaining acceptable values of SAR, which in our case were between 85% and 100% of the maximum allowable. Finally, as the B_1 field varies smoothly over the imaged volume, only modest spatial resolution imaging is required which would permit further reduction of the total acquisition time. These strategies may permit whole-brain B_1 mapping in just a few seconds.

In our implementation of the SS-DAM, the slab-select gradient was not applied during the RF pulse. This avoids loss of signal intensity toward the edges of the prescribed volume (15) as well as other issues (36). However, this may lead to a greater off-resonance frequency effects, and consequent effects on B_1 mapping accuracy (36). While our results indicate that both SS-DAM and DAM provide accurate results in the regime of low-to-moderate off-resonance effects, we have also found that the error in calculated B_1 becomes significant when off-resonance effects are large (Fig. 8), in agreement with previous observations (15). This is expected to be the case in regions with strong susceptibility gradients, including, for example, the regions surrounding the sinuses. Nevertheless, this potential issue can be attenuated using specialized RF pulses (37). In addition, B_0 inhomogeneity leads to image distortion and signal loss, especially with gradient-echo-type sequences. However, this effect may be attenuated using a short TE, as is the case with the SS-DAM. Nevertheless, further work is required to analyze the impact of B_0 inhomogeneity on derived B_1 maps using the SS-DAM.

For simplicity, we analyzed the case when only magnitude images, that is, unsigned signals, are available, and $\theta_1 = \theta_2 = 60^\circ$. While our results indicate that this leads to a large usable range of ζ with a one-to-one relationship between Q and ζ , this one-to-one relationship is not valid outside the interval $0.75 < \zeta < 1.45$ (Fig. 2). This may limit the use of the SS-DAM method for extremely large ζ variations, as may be the case particularly at ultra-high field. While the optimal ζ measurement range can be adjusted through use of different values of θ_1 and θ_2 , these values must be chosen carefully to ensure independence of Q on T_1 (Fig. 2).

Aside from accuracy, advantages of the SS-DAM include rapid data acquisition, simple modeling, use of a conventional MRI sequence, *i.e.* SPGR, available on all preclinical and clinical MRI machines, and rapid processing time. These all suggest the possibility of integrating SS-DAM into routine data processing protocols, including clinical investigations involving, for example, quantitative MRI (1, 4–6, 38, 39), B_1 shimming (40, 41), and image intensity correction (42).

In conclusion, the method presented here, SS-DAM, permits fast, accurate, and practical mapping of the active B_1 field in the clinical setting. Whole brain coverage can be obtained in less than 1 min.

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Figure 1.

a-c) Plot of simulated SPGR signal (Eq. 1) as a function of flip angle (FA), α , for different values of TR/T_1 . b-d) Zoomed view of the signal linearity region centered at $\alpha = 180^\circ$. Note that T_1 has only a rotating effect on S_{SPGR} around $\alpha = 180^\circ$. The first row represents the signed signal values in order to better illustrate the linearity of the SPGR signal, while the second row represents the unsigned signal values used to compute Q .

Figure 2. Plot of simulated $Q = |S_{\text{SPGR}}(\zeta(180^\circ - \theta), \text{TR})| / |S_{\text{SPGR}}(\zeta(180^\circ + \theta), \text{TR})|$ ratio as a function of ζ for different values of TR/T_1 and θ (top row). For better visualization, plots representing zoom on the y-axis were also displayed (bottom row).

Figure 3.

Simulation results for ζ_{est} and relative error (RE) maps as a function of T_1 , for a synthetic brain image. Reference ζ and T_1 maps (left column), and estimated maps, ζ_{est} , using SS-DAM, using four ζ different look-up tables (Q vs. B_1) generated with T_1 values of 500 ms, 1000 ms, 2000 ms or 4000 ms (right column upper row). Results were obtained with $TR = 35$ ms. RE maps and mean and SD RE values, calculated over all brain voxels, are also displayed (right column lower row). The red arrows indicate regions with slightly larger RE. This is due to the fact that the ζ values (i.e. < 0.75) in those regions were outside the usable range of ζ for $\theta = 60^\circ$ (Fig. 2).

Figure 4. Bloch simulation results of the number of RF pulse excitations required to reach the SS (upper row), and the Q factor as a function of ζ (lower row) for different tissue types and transverse magnetization spoiling methods. Zoom on the y-axis of Q vs. ζ are also displayed. Results were obtained with TR = 35 ms.

Figure 5.

Simulation results for ζ_{est} and relative error (RE) maps obtained using the SS-DAM from T_1 -weighted images generated, using Eqs. 2–3, from incomplete magnetization spoiling. T_1 -weighted images were generated from SS signals obtained with crusher gradients and constant-phase pulses or from SS signals obtained with crusher gradients combined with RF phase incrementation using the ζ and T_1 input maps described above. For each spoiling method, ζ maps, RE maps, and corresponding mean and SD RE values were calculated over all brain voxels.

Figure 6.

Simulation results for ζ_{est} and relative error (RE) maps as a function of SNR, for a synthetic brain slice. Reference ζ map (left column), and estimated ζ maps, ζ_{est} , using SS-DAM or DAM, from SPGR T_1 -weighted brain images generated at three different SNRs of 10, 100, and 1000, corresponding to low, clinical, and high SNR regimes, respectively (right column upper row) are shown. For each SNR, RE maps, and mean and SD RE values, calculated over all brain voxels, are also displayed (right column lower row).

Figure 7. Phantom results. DAM and SS-DAM SPGR weighted images and corresponding ζ and RE maps obtained from a cylindrical phantom of water doped with copper sulfate. Results are displayed for two different slices. For each slice, mean and SD RE values are computed.

Figure 8. Phantom results. Percentage deviation in ζ derived from the SS-DAM or DAM at different RF frequency offsets, $\Delta\nu$.

Figure 9. In-vivo results. DAM and SS-DAM SPGR images, and corresponding ζ and RE maps obtained from the brains of two participants. For each slice, mean and SD RE values are computed.

Figure 10.

In-vivo results. ζ maps obtained using SS-DAM and corresponding ζ and RE maps from the brain of the first participant using different imaging acceleration methods. For each method, results are shown for three representative slices. For each slice, mean and SD RE values are computed.