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A generalized slab-wise framework for parallel transmit multiband RF pulse design

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Abstract

Purpose—We propose a new slab-wise framework to design parallel transmit multi-band pulses for volumetric simultaneous multi-slice imaging with a large field of view along the slice direction (FOV_s).

Theory and Methods—The slab-wise framework divides FOV_s into a few contiguous slabs and optimizes pulses for each slab. Effects of relevant design parameters including slab number and transmit B1 (B1+) mapping slice placement were investigated for human brain imaging by designing pulses with global or local SAR control based on electromagnetic simulations of a 7T head RF array. Pulse design using in-vivo B1+ maps was demonstrated and evaluated with Bloch simulations.

Results—RF performance with respect to SAR reduction or B1+ homogenization across the entire human brain improved with increasing slabs; however, this improvement was non-linear and leveled off at ~12 slabs when the slab thickness reduced to ~12 mm. The impact of using different slice placements for B1+ mapping was small.

Conclusion—Compared to slice-wise approaches where each of the many imaging slices requires both B1+ mapping and pulse optimization, the proposed slab-wise design framework is shown to attain comparable RF performance while drastically reducing the number of required pulses; therefore, it can be used to increase time efficiency for B1+ mapping, pulse calculation and sequence preparation.

Keywords

parallel excitation; simultaneous multi-slice imaging; transmit B1 homogenization; high field MRI; multiband RF pulse design

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Supporting Material

Detailed imaging protocols utilized for in vivo B1+ mapping and T1-weighted 3DMPRAGE acquisition.

Introduction

Subsequent to its introduction for accelerating high resolution functional image acquisition in the human brain (1,2), Simultaneous Multi-Band (MB), Multi-Slice (SMS) MR imaging (3) using MB RF pulses has become increasingly popular in the neuroimaging community for functional (fMRI) and diffusion MRI (dMRI) studies. Most notably, the method was adopted in the Human Connectome Project (HCP) (4-10) which is mostly being conducted at 3 Tesla (3T), and, to smaller extent and with more ambitious aims, at 7T (6). However, optimum slice acceleration with SMS/MB imaging for dMRI at both field strengths, but particularly at 7T, is precluded due to transmit B1 (B1+) inhomogeneities and elevated RF power deposition in tissues (i.e., specific absorption rate or SAR). For example, the 3T dMRI protocol of the Washington University-University of Minnesota (WU-Minn) consortium runs with a suboptimal TR of 5.7 s using an MB factor (i.e. slice acceleration factor) of 3 (6). Although much higher MB factors are feasible and employed for gradient-echo-based resting state fMRI (rfMRI) (6), the spin-echo dMRI acquisition is constrained in slice acceleration ultimately because of SAR. Another problem confronted with dMRI is the B1+ inhomogeneities, which result in a non-uniform signal-to-noise ratio (SNR) within slices; the same would be applicable for spin-echo based fMRI (e.g. (11)). Although the problem is particularly pronounced at 7T, it also exists at 3T, where flip angles can differ as much as 30% between the central and peripheral parts of the brain (6).

Previously, we introduced a strategy for parallel transmit (pTx) MB RF pulse design where B1+ non-uniformity can be simultaneously tackled with RF power limitations (12). In this proof of concept study, a few brain slices were optimized in a slice-wise fashion for simultaneous excitation and subsequent imaging at 7T. The results showed that the new slice-wise pTx MB pulses were capable of concurrently reducing B1+ inhomogeneity and constraining total RF power of the pulse (proportional to global SAR) (12) as compared to a single-channel Circularly Polarized (CP) mode application. This slice-wise pTx MB approach in fact provides a generalized framework allowing simultaneous B1+ homogenization and constraint of power parameters including total RF total power (12), peak RF power (13-15), local and global SAR (14,16). It offers several advantages over other existing pulse design methods that can also be employed for high field SMS/MB imaging. Unlike PINS pulse design (17), pTx MB pulses can be used to excite a pre-defined number of slices that cover the desired object and does not require the slice direction to coincide with the restricted dimension of the object (e.g., coronal or sagittal sections in brain imaging) or additional pulses in a spin-echo sequence to impose the restriction. Furthermore, slice-wise pTx MB pulse design allows for simultaneous B1+ homogenization, which is not straightforward for PINS pulses. Unlike the multi-transmit strategy reported by Poser et al. (18) or a similar approach described earlier (19) for MB pulses, the slice-wise pTx MB pulse design does not require any particular coil structure and can be employed even when coil elements do not align with the slice direction, e.g., when exciting axial slices using an RF array with azimuthally distributed elements (12); the slice-wise pTx MB pulse design can in fact be employed to achieve better RF performance using more optimum coil structures where coil elements are distributed along all three spatial dimensions (20).

However, in practice, utilizing this slice-wise approach for a large number of slices to cover a sizeable field of view along the slice direction (FOV_s) becomes difficult within the time constraints of a typical imaging session. This is mainly because the B1+ mapping, RF pulse calculation and pulse sequence preparation together can be very costly in time when they are obtained for each imaging slice. For example, to cover the whole brain (including cerebrum, cerebellum, and brainstem) with a 1.25-mm spatial resolution in the slice direction, ~104 slices would be needed. For 4-fold MB acceleration, this requires a total number of 26 or 52 distinct pTx MB pulses to be maintained in a gradient- or spin-echo sequence, respectively. Furthermore, considering 8-channel transmission, obtaining multi-channel B1+ maps for each imaging slice would require 832 B1+ maps to be calculated and the use of these slice-wise B1+ maps in the subsequent RF shim calculation would require a large-scale optimization problem to be solved, thereby imposing extra time and computational requirements. Using more channels and higher spatial resolutions with more imaging slices would further aggravate the situation. Therefore, seeking alternative pulse design strategies that can avoid the abovementioned time and computational burdens is important in achieving volumetric coverage in pTx SMS/MB imaging.

Traditionally, pTx non-MB pulses for volumetric imaging are designed at a much lower spatial resolution than the actual imaging resolution. For example, in a study demonstrating the usefulness of RF shimming for multi-slab cerebral time of flight (TOF) angiography at 7T (21), we showed that each of the three single-slab pulses can be designed considering only a single 6-mm thick B1+ mapping slice positioned in the middle of the corresponding slab, while providing satisfactory B1+ homogenization across the entire 24-mm thick slab when acquiring high resolution (0.5 mm^3) TOF images. In another study, Malik et al. (22) report that the use of three orthogonal slices for B1+ mapping in axial, coronal and sagittal orientation, may yield reasonable B1+ homogenization across the entire brain when designing spiral non-selective (SPINS) pTx RF pulses for brain imaging at 3T. The success of these previous studies relies on the fact that B1+ field variations occur at fairly low spatial frequencies compared to actual imaging spatial resolution, allowing for satisfactory pulse design based on relatively sparse B1+ maps.

Following this concept, we propose a new *slab-wise* framework for pTx MB pulse design where RF pulses are optimized for a few slabs consisting of sub-groups of imaging slices. The number of pTx MB pulses to be designed is thus determined by the slab number, not by the number of imaging slices. As previously discussed for traditional non-MB pTx pulse design, multi-channel B1+ maps acquired from a reduced number of gapped slices can be utilized to design satisfactory slab-wise pulses. Here we present a thorough investigation of this slab-wise design framework, originally introduced in the format of a conference abstract (16), in the context of whole brain imaging with multiband RF excitation at 7T. RF performance was characterized as a function of key design parameters including the slab number and the slice placement for B1+ mapping. A 7T 16-channel transmit and receive (Tx/Rx) head RF array was simulated to design pulses with either global or local SAR constraint. Additionally, B1+ maps were experimentally measured from human subjects at 7T with a similar 16-channel Tx/Rx array to design pulses with integrated RF power control. In all cases, Bloch simulations were performed to evaluate resulting RF performances.

Theory

Conventional MB RF excitation

In conventional single-channel MB excitation with an MB factor of M , a single MB RF pulse, formed as the sum of M Single-Band (SB) pulses, simultaneously excites M slices. RF pulses executing different spin transformations in a sequence (e.g. excitation, inversion, refocusing, etc.) can be designed as MB pulses based on the same approach. However, for simplicity, only excitation pulses are considered in the following discussion.

The SB pulses of the MB pulse typically have the same base pulse shape (e.g. a sinc shaped pulse), but each modulated with a different temporal phase evolution to have a different frequency about which they achieve excitation of a different slice. These SB pulses are typically chosen such that the M simultaneously excited slices are equidistantly distributed, with the inter-slice distance given by FOV_s divided by M . This leads to M contiguous and equally thick *subsections* along the slice direction, each covered by one SB component of the MB pulse. To cover the entire FOV_s with a *total* of N imaging slices, the central frequency of the MB pulse is stepped for N/M consecutive excitations. As a result, each SB component excites N/M adjacent slices within the respective subsection.

Two-stage slab-wise pTx MB pulse design

In the original slice-wise design (12), each SB component of a pTx MB pulse is specifically optimized for one imaging slice. For N imaging slices, this yields N/M different pTx MB pulses each composed of M differently optimized SB components. In this procedure, the B1+ map within each imaging slice has to be obtained.

In the slab-wise design, each SB component of a pTx MB pulse is optimized simultaneously for multiple imaging slices. This is done by further dividing each of the M subsections into *slabs* and optimizing each SB component for one such slab. The slices within a given slab are then excited sequentially with the same SB component specifically optimized for that *slab*. The scale of optimization procedures to run and the number, K , of different pTx MB pulses to be generated is now determined by the number of slabs per subsection rather than N/M . Furthermore, the B1+ mapping slices do not have to be from the imaging slices.

We outline below the implementation of the slab-wise pulse design in the context of a *pre-calculation* stage where the slabs and B1+ mapping slices are determined, and a *calculation* stage where the complex RF shim values (representing multichannel RF magnitude and phase modulations) for individual SB components are calculated and the final pTx MB pulses assembled.

Pre-calculation stage

For a given MB factor M , the first step in the pre-calculation stage is to specify K , which determines the slab number per subsection as well as the number of different pTx MB pulses to be generated. The product, $K \times M$, then defines the *total* number, N_{slab} , of slabs to be prescribed. In the limit where K reaches N/M , the “slab-wise” design becomes a “slice-wise” design. Fig. 1a illustrates example slab prescriptions for pulse design with $M=3$ targeting

whole brain coverage with axial slices. The left-most figure in Fig.1a identifies the three *subsections* generated by dividing FOV_s by M . When $K=1$ *slabs* and *subsections* are the same (Fig. 1a, left most figure). For $K=2$ and 3, each subsection is further divided into 2 and 3 slabs, respectively, yielding a total number of 6 and 9 slabs. Note that the slab thickness need not be uniform within a subsection, but each subsection has to be divided in the same way; this is because slab thickness is allowed to vary between different pTx MB pulses, but has to be equal for all SB components of a given MB pulse.

The second step in the pre-calculation stage is to determine the placement of the B1+ mapping slice(s) within individual slabs. Various B1+ mapping slice placements may be considered for a given slab configuration. For example, for the same 6-slab configuration (Fig 1a, middle figure) prescribed to design two pTx MB pulses (i.e., $K=2$ and $M=3$), one may consider a single B1+ mapping slice per slab or an arbitrary number of 14 equidistantly placed slices as illustrated from left to right in Fig. 1b. It should be noted that the determination of B1+ mapping slices could in fact precede the prescription of the slab distribution through space.

Calculation stage

After completion of the pre-calculation stage, the calculation stage is conducted to obtain the desired RF shim values, which are typically obtained with optimization procedures that may include a variety of constraints. For example, one may seek RF shim values to minimize RF power or SAR while satisfying predefined excitation fidelity by solving a constrained minimization problem as follows,

$$\hat{\mathbf{w}} = \arg \min_{\mathbf{w}} R(\mathbf{w}), \quad \text{s.t.} \quad RMSE(\mathbf{w}) \leq \epsilon \quad [1]$$

where $\mathbf{w}$ is a complex-valued vector concatenating the RF shim values for individual pTx MB pulses; $R(\mathbf{w})$ is the cost function used to minimize power or SAR; $RMSE(\mathbf{w})$ represents the root mean square excitation error; the number ϵ is used to constrain the excitation error. The excitation error can be defined in a magnitude least square (MLS) sense (23) by

$$RMSE(\mathbf{w}) = \|\mathbf{A}\mathbf{w} - \mathbf{d}\|_2 / \sqrt{N_{ROI}} \quad [2]$$

Where $\mathbf{A}$ is the complex-valued system matrix involving the multi-channel B1+ maps, $\mathbf{d}$ a real valued vector representing the desired excitation, and N_{ROI} the number of voxels in the region of interest (ROI). Since a different set of RF shim values is sought for each pTx MB pulse, the system matrix $\mathbf{A}$ has a characteristic block-diagonal structure, i.e.,

$$\mathbf{A} = \begin{bmatrix} \mathbf{A}_1 & \cdots & 0 \\ \vdots & \ddots & \vdots \\ 0 & \cdots & \mathbf{A}_K \end{bmatrix}.$$

Correspondingly, the vectors $\mathbf{w}$ and $\mathbf{d}$ are written as

$$\mathbf{w} = \begin{bmatrix} w_1 \\ w_2 \\ \vdots \\ w_K \end{bmatrix} \quad \text{and} \quad \mathbf{d} = \begin{bmatrix} d_1 \\ d_2 \\ \vdots \\ d_K \end{bmatrix}.$$

In the case of band-specific pulse design where each SB component of a given pTx MB pulse has a different RF shim set, the matrix $\mathbf{A}_k$ is also constructed to be block-diagonal

$$\mathbf{A}_k = \begin{bmatrix} \mathbf{A}_{k,1} & \cdots & 0 \\ \vdots & \ddots & \vdots \\ 0 & \cdots & \mathbf{A}_{k,M} \end{bmatrix}.$$

Correspondingly, vectors $\mathbf{w}_k$ and $\mathbf{d}_k$ further read

$$\mathbf{w}_k = \begin{bmatrix} w_{k,1} \\ w_{k,2} \\ \vdots \\ w_{k,M} \end{bmatrix} \quad \text{and} \quad \mathbf{d}_k = \begin{bmatrix} d_{k,1} \\ d_{k,2} \\ \vdots \\ d_{k,M} \end{bmatrix}$$

For C -channel transmit, the individual diagonal element matrices of matrix $\mathbf{A}_k$ are given

$$\mathbf{A}_{k,m} = i\gamma \cdot f \left[\mathbf{A}_{k,m,1}, \mathbf{A}_{k,m,2}, \cdots, \mathbf{A}_{k,m,C} \right] \quad \text{where} \quad \mathbf{A}_{k,m,c} = \begin{bmatrix} \mathbf{a}_{k,m,c,1} \\ \mathbf{a}_{k,m,c,2} \\ \vdots \\ \mathbf{a}_{k,m,c,L}(k,m) \end{bmatrix}.$$

$$\mathbf{w}_{k,m} = \begin{bmatrix} w_{k,m,1} \\ w_{k,m,2} \\ \vdots \\ w_{k,m,C} \end{bmatrix} \quad \text{and} \quad \mathbf{d}_{k,m} = \begin{bmatrix} d_{k,m,1} \\ d_{k,m,2} \\ \vdots \\ d_{k,m,L}(k,m) \end{bmatrix}.$$

Correspondingly,

Here k , m , c and l count the pTx MB pulses, the , the transmit channels and the B1+ mapping slices, respectively; γ is the gyromagnetic ratio, f the time integration of the base RF pulse shape and $i = \sqrt{-1}$. For designing band m of the k -th MB pulse, $L^{(k,m)}$ denotes the number of B1+ mapping slices, $\mathbf{a}_{k,m,c,l} (l = 1, 2, \dots, L^{(k,m)})$ is a vector concatenating the B1+ map of channel c obtained within the l -th B1+ mapping slice, $\mathbf{w}_{(k,m,c)}$ is the RF shim value to be calculated for channel c , and $\mathbf{d}_{k,m,l}$ is a vector representing the target excitation predefined within the l -th B1+ mapping slice. It should be noted that the terms $\mathbf{A}_{k,m}$, $\mathbf{w}_{k,m}$ and $\mathbf{d}_{k,m}$ described above are the same as those used in conventional RF shimming to calculate RF shim values for a single slice ($L = 1$) (24) or simultaneously for multiple slices ($L > 1$) (25).

The cost function in Eq. [1] can be used to control a RF power related quantity such as global SAR, local SAR or integrated RF power. For global SAR control, it can be written as

$R(\mathbf{w}) = \sum_{k=1}^K \sum_{m=1}^M \mathbf{w}_{(k,m)}^H \mathbf{Q} \mathbf{w}_{(k,m)}$ where the superscript H denotes complex conjugate transpose, and $\mathbf{Q}$ is the global SAR matrix (26). For local SAR control, the cost function can

be defined as $R = \max_{\{n=1,2,\dots,N_{VOP}\}} \sum_{k=1}^K \sum_{m=1}^M \mathbf{w}_{(k,m)}^H \mathbf{S}_n \mathbf{w}_{(k,m)}$ where N_{VOP} is the number of virtual observation points (VOP's) that can be created using the model compression method as described by Eichfelder and Gebhardt (27), and $\mathbf{S}_n$ is the n -th VOP. For integrated power control, $R(\mathbf{w})$ can simply be given by $R(\mathbf{w}) = \mathbf{w}^H \mathbf{w}$. In each case, Eq. [1] can be cast into an equivalent regularized MLS problem and solved using the algorithms described by Lee et al. (28) and Sbrizzi et al. (29).

Once the RF shim values are obtained, the channel- c component of the k -th pTx MB pulse

can be assembled as $p_{k,c}(t) = \sum_{m=1}^M w_{(k,m,c)} \cdot a_m(t)$. Here $a_m(t)$ denotes the m -th unshimmed SB pulse waveform as used in the conventional single-channel MB excitation and is given by $b(t) \cdot \exp[-i\gamma G \cdot (m-1)z_0 \cdot t]$ where $b(t)$ is the base pulse shape, G the maximum amplitude of the slice-selective gradient and z_0 the inter-slice distance between the SB pulses.

Methods

Electromagnetic (EM) field maps of a 16 channel Tx/Rx 7T head RF array employed in our laboratory (30,31) were simulated for pulse design. The RF array was modeled using the XFDTD software (Remcom, USA) and was loaded with a generic human head and shoulder model with a medium head size. The tissue model consisted of 680391 tissue voxels and was defined with spatial resolutions of 2, 2 and 2.5 mm in the X , Y and Z axes, respectively. The brain region spanned 144 mm in the Y axis with 72 coronal slices. The RF array consisted of 16 microstrip elements (30,31), each measuring 180 mm in length and 12 mm in width. The 16 elements were arranged azimuthally and evenly on a single 270-mm-diameter ring and were positioned inside a coaxial RF shield made of a cylindrical copper sheet (320 mm in diameter and 210 mm in length) which also provided grounding. The dielectric material used in between the coil elements and the ground plane was air. The 16-channel EM maps within the entire tissue model were simulated by driving one element at a time with all other present. Fig. 2 illustrates the EM modeling of the array along with the 16-channel $|B_1|$ maps in an axial slice.

These simulated EM fields were employed to design pulses with global or local SAR control. The local SAR matrix per tissue voxel was computed based on an approach proposed by Caputa et al. (32) to ensure exact 10 g of cubic volume for SAR averaging; the region growth algorithm involved in this computation was one similar to that described by Zelinski et al. (33). The resulting 680391 local SAR matrices were used to create 55 VOP's which were utilized in the local SAR controlled pulse design. The base pulse shape utilized to assemble final pTx MB pulses was a 1-ms filtered sinc pulse with a bandwidth-time-product of 6. All pulses were designed for a nominal flip angle of 10° and for excitation of

coronal slices. Only brain voxels were included in the ROI. All calculations except for EM simulations were performed in Matlab (Mathworks, USA).

Pulse design with global SAR control

We started by designing global SAR controlled pulses to investigate how RF performance of the slab-wise design would be affected by relevant design parameters such as total slab number and B1+ mapping slice placement.

To investigate the impact of slab number, we conducted pulse designs for $M=3$ and $M=4$. For each MB factor, N_{slab} was increased by incrementing K . To keep the slab number as the only variable that may influence the resulting RF performance, all 72 brain slices were employed as B1+ mapping slices; additionally, for a given N_{slab} , slab thickness variation was *not* allowed within a subsection. Given that total number of slices divided by N_{slab} must be an integer and that N_{slab} must also be an integer multiple of M (since $N_{\text{slab}}=K \times M$), 72 slices resulted in eight possible slab configurations for $M=3$ (i.e., $N_{\text{slab}} = 3, 6, 9, 12, 18, 24, 36$ and 72) and six for $M=4$ (i.e., $N_{\text{slab}} = 4, 8, 12, 24, 36$ and 72), each containing $72/N_{\text{slab}}$ number of B1+ mapping slices per slab.

After studying the impact of slab numbers, we investigated the effect of B1+ mapping slice placement. This was done by using the same slab configuration but with various numbers and positions of B1+ mapping slices. Specifically, we considered a pulse design for $M=4$ and decided to use the 8- and 12-slab configurations previously employed in investigating the impact of slab numbers with constant (i.e. uniform) slab thicknesses, thus leaving B1+ mapping slice placement as the only variable in pulse design. For each slab configuration, three strategies for choosing B1+ mapping slices from the 72 available brain slices were investigated, including the use of 1) all 72 slices, 2) equidistantly placed $3N_{\text{slab}}$ number of B1+ mapping slices across the brain (i.e., three slices per slab), and 3) a single slice per slab. The RF performance of each strategy was evaluated by considering B1+ distributions across all 72 brain slices.

Pulse design with local SAR control

We also investigated the impact of slab numbers in *local* SAR controlled pulse design since local SAR raises greater safety concerns when using pTx for ultra-high field MRI. Similar to the global SAR controlled pulse design study, we considered $M=3$ and $M=4$; for each MB factor, the total slab number utilized in pulse calculation was varied by increasing K , and the slab thickness was kept constant. Because using all 72 brain slices as B1+ mapping slices was found unnecessary for optimum RF performance for global SAR controlled pulses, we decided to only use 24 B1+ mapping slices that were evenly distributed across the brain, thereby reducing the computational burden in pulse calculation. This yielded four slab configurations for $M=3$ (i.e., $N_{\text{slab}} = 3, 6, 12$ and 24) and for $M=4$ (i.e., $N_{\text{slab}} = 4, 8, 12$ and 24), each with $(24/N_{\text{slab}})$ number of B1+ mapping slices per slab.

RF performance evaluation

The RF performance was evaluated by varying ε in Eq. [1] to create L-curves to quantify tradeoffs between the controlled SAR quantity (i.e., global SAR or peak 10-g local SAR)

and excitation error. For comparison, an RF phase shimming targeting a CP-like mode B1+ distribution in the entire brain (34) was conducted to mimic a single-channel transmission. In order to appreciate the potential gains in B1+ homogeneities with pTx MB pulses, all L-curves were plotted using the normalized RMSE (NRMSE) calculated by $\text{NRMSE} = \text{RMSE} / \text{RMSE}_{\text{CP}}$ (where RMSE and RMSE_{CP} are the excitation errors for pTx MB and the CP mode pulses, respectively). The RMSE_{CP} value was calculated to be 0.042, considering B1+ distributions within all 72 brain slices. Unless otherwise noted, RMSE values were calculated based on the resulting B1+ distributions only within the B1+ mapping slices. SAR quantities were calculated for an excitation of 144 slices within a second, assuming whole brain single-shot 2D-EPI with a 1-sec volume TR and 1-mm slice thickness. Peak 10-g local SAR was evaluated by considering all local SAR matrices.

Demonstration using in vivo B1+ maps

To demonstrate the feasibility of the proposed slab-wise pulse design framework in a more real situation, we acquired 16-channel 24-coronal-slice B1+ maps in the human brain at 7T and utilized these maps to design slab-wise pTx MB pulses. The 24 B1+ mapping slices were 4-mm thick and equidistantly placed across the brain region. A fast hybrid technique (35,36) combining a high flip angle 3D AFI (37) with a series of small flip angle multi-slice gradient echo (GRE) images was employed. These 24 B1+ maps were employed to design pulses for $M=2$, and allowed the number of pTx MB pulses to be $K=1, 2, 3, 4, 6$ and 12 , which corresponded to $N_{\text{slab}}=2, 4, 6, 8, 12$ and 24 , respectively, with $24/N_{\text{slab}}$ B1+ mapping slices per slab. Only integrated RF power was controlled (i.e., using $R(\mathbf{w}) = \mathbf{w}^H \mathbf{w}$ in Eq. [1]) since controlling integrated power is most commonly used for pulse design in pTx experiments (12,15,23,25). The performance of the resulting pulses with respect to whole brain excitation was evaluated by Bloch simulations using a different set of contiguous B1+ maps covering the entire brain and acquired with 2-mm thickness. The binary masks defining the brain region in the pulse design and in the RF performance evaluation were created in FreeSurfer (<http://surfer.nmr.mgh.harvard.edu/>) using a hybrid skull stripping approach (38); the T1 weighted image utilized for this procedure was derived by dividing a standard 3D MPRAGE image obtained with 1-mm isotropic spatial resolution by a matched proton density image so as to remove the biases arising from proton density, T2* contrast and RF coil receive B1 sensitivity (39).

All experiments were performed on a 7T scanner (Siemens, Erlangen, Germany) driven by a customized 16-channel transmit system equipped with 1 kW RF amplifier per channel. A 16-element RF transceiver array (31) similar to the one used in the simulation study was employed for both RF transmission and reception. Details of the imaging protocols utilized for B1+ mapping and 3D MPRAGE acquisitions are reported in Supporting Material. Healthy volunteers who signed a written consent approved by local Institutional Review Board were scanned.

Results

Global SAR controlled pulse design

For both $M=3$ and $M=4$, RF performance in terms of SAR reduction and/or B1+ homogenization improved with increasing slabs (Fig. 3a and 3c, for $M=3$ and 4, respectively); however, this improvement was non-linear and leveled off at ~ 12 slabs which was significantly less than the number of imaging slices (i.e., before the slice-wise design was reached at $N_{\text{slab}}=72$). Further plotting global SAR versus the utilized slab number at a given NRMSE exhibited an L-shaped curve (Fig. 3b and 3d, for $M=3$ and 4, respectively).

Fig. 4 depicts the effect of B1+ mapping slice placement when designing pulses for $M=4$. As can be seen from the L-curves (Fig. 4b and 4d) generated for the three representative B1+ mapping slice placement strategies investigated for $K=2$ (Fig. 4a) and $K=3$ (Fig. 4c), using gapped B1+ mapping slices appeared to have limited impact and slightly degraded the RF performance as compared to using all 72 brain slices as B1+ mapping slices. Notably, using 3 evenly placed slices per slab for B1+ mapping yielded nearly the same RF performance as using all 72 slices for both $K=2$ and $K=3$. For $K=3$, where the slabs are thinner, even a single B1+ mapping slice per slab was sufficient to produce as good RF performance as using all 72 slices.

Local SAR controlled pulse design

Similarly to global-SAR-controlled case, RF performance with local-SAR constraint improved with increasing number of slabs, exhibiting a nonlinear behavior that plateaued at ~ 12 slabs when the slab thickness reduced to ~ 12 mm (Fig. 5).

To investigate how local SAR was impacted for a given M by designing more pTx MB pulses with a larger number of slabs, we chose the pulse design for $M=4$ and compared local SAR distribution at a constant NRMSE of 0.57 when designing a single ($K=1$) vs. three ($K=3$) pTx MB pulses (Fig. 6). For $K=3$, each of the four subsections (as determined by $M=4$) was further divided into 3 slabs for pulse calculation, resulting in 12 slabs in total (as indicated by yellow bars in Fig. 6a (top)) spanning the entire brain in the slice direction. When the 3 pTx MB pulses had finished exciting all image slices within the four slabs that they were designated to excite, the 3 resulting 10-g local SAR distributions were different than each other (Fig. 6a, bottom row), and more importantly, they presented lower hotspots than those observed in the case of $K=1$ (Fig. 6a, middle row) when the single pTx MB pulse was repeatedly applied to excite the same 12 slabs. As a result, the overall accumulated SAR corresponding to whole brain excitation was reduced by 18% for $K=3$ (Fig. 6b). Further checking the SAR contributions from each individual SB component (Fig. 6c) unveiled that some of the 12 SB SAR distributions exhibited a lower peak amplitude in the SAR hotspot for $K=3$ than for $K=1$ (e.g., when exciting the most anterior slab).

Fig. 7 presents the whole brain B1+ distributions across all 72 brain slices at NRMSE=0.57 for the pulses designed with $M=4$ and $K=3$. Although the pulses were only optimized for the 24 gapped B1+ mapping slices, consistent excitation fidelities were observed in non B1+ mapping slices, yielding satisfactory B1+ distribution across the entire brain with NRMSE calculated to be 0.58. The corresponding coefficient of variation (CV), defined as the

standard deviation of the B1+ distributions divided by the mean (i.e., $\text{std}(|B1+|)/\text{mean}(|B1+|)$), was calculated to be 13%.

Pulse design using in vivo B1+ maps

As was observed using simulated EM field maps, the RF performance of the pulses designed using measured B1+ maps improved with increasing slabs (Fig. 8). The plots of integrated power versus total slab numbers at a given NRMSE further presented nonlinear behavior with an L-shaped pattern where the integrated power started to plateau at ~12 slabs (Fig. 8b).

Additionally, the use of 24 gapped B1+ mapping slices for pulse design was sufficient to accomplish the targeted excitation fidelity across the entire brain. This can be seen in Fig. 9 displaying the resulting B1+ distributions for the pulse designed using 12 slabs and $\varepsilon = 0.024$. Although optimized only to the 24 gapped B1+ mapping slices, the pulse was able to produce comparable B1+ homogeneities across the entire brain which in this representative subject consisted of 84 contiguous 2-mm thick coronal slices.

Discussion

We have demonstrated a new slab-wise design framework that can be used to facilitate the application of pTx MB pulses for volumetric SMS/MB imaging with large FOV_s. Within this slab-wise framework, the calculation of RF shim values is performed based on fewer slabs as opposed to each imaging slice. Additionally, multi-channel B1+ mapping can conveniently be limited to a reduced number of gapped slices.

An important feature of the slab-wise design framework is that different design parameters can be independently adjusted for a given MB factor. In the current study, two design parameters were explored: the slab number and the B1+ mapping slice placement. To investigate their impact on RF performance, we designed pulses where a single parameter was varied while the other remained unchanged. As expected, our results, obtained using global or local SAR regularized pulse design targeting whole brain coverage at 7T, indicated that increasing the slab number initially improved whole brain B1+ homogenization or SAR reduction; however, this improvement in RF performance plateaued at a slab number significantly smaller than the number of imaging slices (Figs. 3 and 5). The effects of B1+ mapping slice placement were found small especially when a 12-slab configuration was utilized for the pulse design (Figs. 4d).

The proposed slab-wise pulse design framework is anticipated to play a critical role in pushing toward volumetric pTx SMS/MB imaging with large FOV_s and high spatial resolutions. Without this framework, the number of pTx MB pulses to be designed is given by the total number of imaging slices divided by the MB factor; the slab-wise pulse design reduces this substantially without a significant compromise on RF performance. For example, when considering pulse design for 4-fold MB acceleration and using a gradient-echo sequence to image the entire brain with coronal slices, our results based on EM simulations of a 7T 16-channel head array show that one can almost accomplish the best achievable RF performance by only designing 3 slab-wise pTx MB pulses using 12 slabs (as

opposed to 18 or even 36 slice-wise pTx MB pulses if 2-mm or 1-mm isotropic resolutions are desired, respectively).

Additionally, slab-wise pulse design for the human brain did not require slice-wise multi-channel B1+ mapping. Our results suggest that RF performance can be well preserved when using only B1+ information obtained from a small number of gapped slices. Consequently, using the slab-wise pulse design will largely reduce the total time needed for B1+ mapping, pulse calculation, and sequence preparation. Further quantitative analyses using our pulse designer to design integrated power controlled pulses for an MB factor of 4 reveal that the computation of RF shim values for 3 slab-wise pTx MB pulses using 24 gapped B1+ mapping slices was 4.7-fold faster than the design of 21 slice-wise pulses with 84 B1+ mapping slices. Furthermore, tested on our 7T MR scanner using a customized pTx MB pulse sequence, the preparation was accelerated by a factor of 3 when applying the 3 slab-wise pulses. Because of its time efficiency, the use of the slab-wise design framework is expected to make volumetric pTx SMS/MB imaging with large FOV_s more feasible within clinical time constraints.

The choice of slab number is important in the proposed slab-wise pulse design framework. Using a too small slab number results in significantly degraded RF performance, whereas using a too large slab number decreases the time efficiency in pulse design and sequence preparation but only provides little or even no improvement in RF performance. The L-shaped curve characterizing the SAR behaviors relative to the slab number at fixed excitation fidelity (e.g., Figs. 3 and 5) suggests that the slab numbers near the corner of the L-curve can be used to provide good balance between SAR penalty and the number of desired pTx MB pulses. In practice, coil-specific L-curves can be created to help the user choose the appropriate slab number for the coil and the target geometry under consideration. This can be done based on the corresponding EM field simulations that are always available for the SAR computation necessary for following safety guidelines. Alternatively, coil-specific L-curves may be created based on experimentally acquired B1+ maps. In this case, B1+ maps within a few gapped slices can be acquired and utilized to reconstruct the B1+ profiles over the full ROI using certain interpolation or fitting algorithms (e.g., through a model-based approach (40)). Otherwise, a sufficient number of B1+ mapping slices are desirable in order to minimize the inaccuracy when only using the B1+ mapping slices to evaluate the excitation fidelity across the entire ROI. For brain imaging at 7T using an RF array with 16 azimuthally distributed coil elements, our results (Figs. 8 and 9) suggest that the use of 24 equidistantly placed B1+ mapping slices is likely to suffice in this regard when exciting coronal slices. Additional simulation studies for other slice directions also suggest that using 24 equidistant B1+ mapping slices may be sufficient in evaluating the whole brain excitation fidelity when designing pulses to excite axial or sagittal slices. For lower magnetic fields, where the B1+ variation over the human brain is less pronounced, fewer slices for B1+ mapping and fewer slabs for pulse optimization may be adequate. Conversely, going to higher static magnetic fields will likely require larger number of slabs and B1+ mapping slices.

The choice of the slab number may also be automated as part of the pulse design procedure. For this, one can first predefine the B1+ mapping slices and identify the possible slab

numbers, i.e. numbers that are simultaneously integer multiples of the MB factor and integer divisors of the number of B1+ mapping slices. For example, using 24 B1+ mapping slices and considering pulse design with $M=4$, the possible slab numbers would be $N_{\text{slab}}=4, 8, 12$ and 24, corresponding to $K=1, 2, 3$ and 6, respectively. The optimum slab number for a given RMSE constraint can then be found from the possible slab numbers by including an outermost loop in the pulse design procedure to repeatedly solve Eq. [1] with increasing slabs until the incremental improvement in RF power or SAR is within a predefined value. A generic algorithm for this is presented in Table 1.

Although in the current study the pulse design was formulated to minimize a single power related quantity while satisfying certain excitation fidelity, the proposed slab-wise design framework can be combined with other pulse design formulations. For example, it can be employed to formulate a constrained minimization that seeks to minimize the excitation error subject to strict constraints simultaneously, for example constraining integrated and peak RF power as well as global and local SAR (41,42) or absolute temperature (43), to produce pTx MB pulses that simultaneously satisfy the RF technical and safety limits (14).

The slab-wise design framework is fully compatible for designing multi-spoke MB pulses (14,15). In the case where simultaneous spoke placement optimization is desired for better RF performance (15), the effectiveness of the slab-wise framework in reducing the pulse computation time is expected to be even more significant, given the substantial additional computation burden required to find the optimum k-space placement of individual spokes. Main field inhomogeneity correction necessary for in vivo multi-spoke pulse design (24,25) can also be incorporated. Finally, the proposed slab-wise pTx MB pulse design framework is fully compatible with the time-shifting strategy (44) described earlier for overcoming peak power limitations in spin-echo sequences.

It should be noted that the proposed slab-wise design framework is applicable for a band-joint pTx MB pulse design (16) where for each pTx MB pulse a common RF shim set is calculated and applied for all the M constituting SB components (12,15,16). The advantage of such band-joint pulse design against the band-specific pulse design is that it does not require the complex and expensive full pTx hardware capable of synthesizing a channel-specific RF waveform for each individual channel; rather it can be achieved using a much simpler and cheaper hardware setup with an RF phase and gain controller, and thus can serve as a useful tool for B1+ homogenization when only this limited pTx capability is available. However, because of the M -fold decrease in the degrees of freedom in the pulse optimization procedure, the band-joint design is expected to provide inferior RF performance as compared to the band-specific design.

Conclusions

We have presented and described a new slab-wise frame-work for pTx MB pulse design, which is aimed at making volumetric pTx SMS/MB imaging with large FOV_s more feasible in real applications. Compared to the slice-wise approach, this new slab-wise design framework can be employed to attain comparable RF performance while only requiring a small number of pTx MB pulses to be designed. Additionally, it does not require multi-

channel B1+ maps to be acquired from each imaging slice. For the human head, we specifically demonstrate that using ~12 slabs with 1~2 B1+ mapping slices per slab are likely to be adequate when using a transmit coil with 16 azimuthally distributed elements. Therefore, the use of the slab-wise framework is expected to drastically improve the time efficiency in B1+ mapping, pulse calculation, and sequence preparation.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIG. 1.

(a) Examples for slab prescriptions when designing pTx MB pulses with an MB acceleration factor of 3 (i.e., $M=3$) to excite axial slices to cover the entire brain. Possible layout of the slab arrangement is depicted for one (left), two (middle) and three (right) pTx MB pulses, leading to a total number (N_{slab}) of 3, 6 and 9 slabs, respectively. Note that it is only necessary to ensure a constant slab thickness for those corresponding to a same pTx MB pulse as shown with the color scheme; the regions along the slice direction covered by the SB components of a given pTx MB pulse are identified using the same color. (b) Example B1+ mapping slice placements when using the same 6 slabs to design two pTx MB pulses for $M=3$.

FIG. 2. Electromagnetic modeling (a) of the 7T 16-channel head array considered in the current study and the B1+ magnitude maps (b) of the 16 individual coil elements within an axial view.

FIG. 3.

Impact of the total slab numbers in slabwise pTx MB pulse design with global SAR control for human head imaging. All pulses were designed in the coronal orientation. Two MB factors were considered: $M=3$ (a, b) and $M=4$ (c, d). For each MB factor, the total number of slabs, N_{slab} , utilized in pulse calculation was varied by incrementing the number, K , of desired pTx MB pulses. In each case, the slab thickness was constant across all slabs and all of the 72 brain coronal slices available in the tissue model were employed as B1+ mapping slices. The corresponding L-curves quantifying tradeoffs between normalized excitation error (NRMSE) with respect to the CP mode and the resulting global SAR are shown in (a) and (c). To better visualize the impact of N_{slab} on pulse performance, the plots of global SAR as a function of N_{slab} at a given NRMSE are presented in (b) and (d) for four different NRMSE values.

FIG. 4.

Impact of B1+ mapping slice placement in slabwise pTx MB pulse design for human head imaging. All pulses were designed in the coronal orientation. For this, MB4 pulse design (i.e., $M=4$) with global SAR control was conducted for $K=2$ (a, b) and $K=3$ (c, d), which led to an 8-slab ($N_{\text{slab}}=8$) and a 12-slab ($N_{\text{slab}}=12$) configuration, respectively. For each slab configuration, the slab thickness was constant, but three B1+ mapping slice placements (a, c) were investigated including the use of: 1) all 72 brain slices, 2) equidistantly placed ($3 \times N_{\text{slab}}$) number of slices across the brain (i.e., three slices per slab), and 3) single slice per slab. The x and y coordinates denote the slice position relative to the total 72 brain slices and to all of the chosen B1+ mapping slices, respectively (e.g., a B1+ mapping slice at slice index 14 and slice number 2 is the 14th brain slice chosen as the 2nd B1+ mapping slice). RF performance for each placement was evaluated by generating the L-curve (b, d) for which the NRMSE values were computed based on the resulting B1+ distributions within all 72 brain slices, not just those within the B1+ mapping slices that had been utilized in the pulse calculation.

FIG. 5.

Investigation of impact of total slab numbers when peak 10 g local SAR is controlled in slabwise pTx MB pulse design. All pulses were designed in the coronal orientation. Two MB factors were considered: $M=3$ (a, b) and $M=4$ (c, d). For each MB factor, the total slab number (N_{slab}) utilized in pulse calculation was varied by incrementing the number, K , of desired pTx MB pulses. In all cases, the slab thickness was constant across all slabs and the same B1+ mapping slice placement consisting of 24 evenly distributed slices across the brain was utilized. The corresponding L-curves quantifying tradeoffs between normalized excitation error (NRMSE) with respect to the CP mode and the resulting peak 10 g local SAR are shown in (a) and (c). Similarly to those in global SAR controlled pulse design, the plots of peak 10 g SAR vs N_{slab} at fixed NRMSE's as shown in (b,d) also exhibited an L shape. Note that local SAR started to plateau when the slab thickness reduced to ~ 12 mm.

FIG. 6.

Comparison of local SAR distributions at a constant NRMSE of 0.57 for a single ($K=1$) vs. three ($K=3$) pTx MB pulses designed to excite coronal slices with $M=4$ and peak 10 g SAR control. Maximum intensity projection (MIP) maps of 10 g SAR arising from each pTx MB pulse are shown in (a), and the MIP of accumulated 10 g SAR after exciting the entire brain in (b). The MIP maps of 10 g SAR for each individual single-band (SB) component are presented in (c) where the first number in the parenthesis counts the pTx MB pulses and the second number counts the SB component within a same pTx MB pulse; e.g. (2, 3) means the 3rd SB component of the 2nd pTx MB pulse. Note that designing a larger number of pTx MB pulses (with a larger N_{slab}) reduced peak local SAR of individual SB components, thereby leading to a lower cumulative maximum.

FIG. 7.

Whole brain flip angle distributions across all of the 72 brain coronal slices at NRMSE=0.57 for the pulses designed to excite coronal slices with $M=4$ and $K=3$ using peak 10 g SAR control. Each row displays the resulting flip angle distributions for the six brain slices within one of the 12 slabs where the brain regions are highlighted by gray lines. The 24 gapped B1+ mapping slices (i.e., 2 slices per slab) utilized in pulse design are indicated by orange boxes. Although the pulses were only optimized for the B1+ mapping slices, consistent excitation fidelities were observed in non B1+ mapping slices.

FIG. 8.

Impact of total slab numbers when designing slab-wise pTx MB pulses in the coronal orientation with in vivo B1+ maps acquired in the human brain at 7T. Integrated power controlled pulses were designed for $M=2$ and $K=1, 2, 3, 4, 6$ and 12 , corresponding to $N_{\text{slab}}=2, 4, 6, 8, 12$ and 24 slabs, respectively. For each K , The same 24 B1+ mapping slices evenly distributed across the brain were utilized. The corresponding L-curves quantifying tradeoffs between normalized RMSE (NRMSE) with respect to the CP mode and the resulting integrated power are shown in (a) and the plots of integrated power vs N_{slab} for different NRMSE values are shown in (b). The RMSE of the CP mode utilized in the normalization was estimated to be 0.043 by considering the B1+ distributions within the 24 B1+ mapping slices.

FIG. 9.

Simulated flip angle (FA) maps in the brain at 7T. The FA maps within the 24 gapped B1+ mapping slices are shown for the CP mode (a) and pTx MB pulses designed with 12 slabs and corresponding to NRMSE=0.56 (b). The region of interest considered for pulse design is indicated by the black curves. (c) Corresponding FA maps across the whole brain. For this representative subject, the 2-mm thick multichannel multislice B1+ maps were acquired within 84 contiguous coronal slices covering the entire brain; the NRMSE within the brain region as indicated by the black curves was calculated to be 0.58 and the coefficient of variation (i.e. $\text{std}(|B1+|)/\text{mean}(|B1+|)$) was 14%.

Table 1

A generic algorithm to optimize the slab number in the slab-wise pTx MB pulse design when using 24 B1+ mapping slices and Eq. [1] to design pulses for $M=4$ and for a constant excitation error ε

1. Find possible slab numbers, i.e., $N_{\text{slab}} \in N = \{4, 8, 12, 24\}$.
2. Start with the least slab number in N .
3. repeat
4. Solve Eq. [1] for $\mathbf{w}$.
5. Evaluate cost, i.e. $R(\mathbf{w})$.
6. Choose next larger slab number in N .
7. until The difference between present and previous cost values is less than a predefined value (e.g., 8% of the previous cost), or the greatest slab number in N is reached.

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