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Automated Slice-Specific Simultaneous Z-Shim Method for Reducing B1 Inhomogeneity and Susceptibility-Induced Signal Loss with Parallel Transmission at 3T

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Abstract

Purpose—Through-plane susceptibility-induced signal loss in gradient recalled echo (GRE)-based sequences can considerably impair both the clinical diagnosis and functional analysis of certain brain areas. In this work, a fully automated simultaneous z-shim approach is proposed on the basis of parallel transmit (pTX) to reduce those signal dropouts at 3T.

Theory and Methods—The approach uses coil-specific time-delayed excitations to impose a z-shim phase. It was extended toward B1 inhomogeneity mitigation and adequate slice-specific signal-dephasing cancellation on the basis of the prevailing B0 and B1 spatial information. Local signal recovery level and image quality preservation were analyzed using multi-slice FLASH experiments in humans and compared to the standard excitation. Spatial blood-oxygen-level-dependent (BOLD) activation coverage was further compared in breath-hold functional MRI.

Results—The pTX z-shim approach recovered approximately 47% of brain areas affected by signal loss in standard excitation images across all subjects. At the same time, B1 shading effects could be substantially reduced. In these areas, BOLD activation coverage could be also increased by approximately 57%.

Conclusion—The proposed fully automated pTX z-shim method enables time-efficient and robust signal recovery in GRE-based sequences on a clinical scanner using two standard whole-body transmit coils.

Keywords

parallel transmission; susceptibility artifacts; z-shim; functional MRI; brain imaging

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INTRODUCTION

In high-field MR imaging, the susceptibility-induced signal loss in T2*-weighted images is still a major problem in many clinically important gradient recalled echo (GRE)-based sequences such as FLASH and susceptibility-weighted imaging. In the human brain, prominent signal losses mainly occur in the frontal orbital and inferior temporal cortex and can seriously impede the diagnosis of, for example, stroke and hemorrhage. Longer echo times (TEs) as required in BOLD functional MRI (fMRI) studies impair the effect, which is particularly caused by the B0 field variations along the slice profile (i.e., through-plane field components).

To address this problem, several techniques have been proposed in the past, but most of them are accompanied by other drawbacks; therefore, they were not practical for clinical routine. Z-shim methods (1–5) have been shown to tackle the through-plane magnetic field variations by applying different gradient z-lobes after the excitation pulse, but they require multiple subimages and can significantly increase scan time. Protocol parameters can be optimized (6,7) for reducing the signal loss, but they come along with decreased coverage and longer pulse repetition time (TR). With additional B0 shim coils (8,9) or diamagnetic shim materials (10–12), the local field inhomogeneity, and thus the signal loss, can be mitigated, but they can increase manual efforts (10–12), discomfort of the patient (9–11), or the need for additional specialized hardware (8). An alternative solution includes custom-designed, three-dimensionally tailored (13–15), and spectral-spatial (16–19) radio frequency (RF) pulses, which have been shown to be promising to compensate for the through-plane susceptibility-induced signal loss without additional hardware or major imaging constraints. However, RF pulse durations can be inefficiently long and increase TE/TR. They can be shortened in combination with the parallel transmit (pTX) technology (13). Nevertheless, these RF pulse design approaches show generally high computational complexity; therefore, they have inadequate computation times for clinical multi-slice imaging. Moreover, the spatial selectivity along the slice direction (i.e., slice selectivity) is not inherently provided but has to be fully optimized. Hence, the RF pulse computation relies on the proper sampling density and coverage of the RF design grid to enable an accurate definition of the slice profile without aliasing effects. The custom designed RF pulses are also difficult to realize within RF hardware and specific absorption rate (SAR) limits. Critical peak voltages occur already at moderate flip angle (FA) levels around 30 degrees and limit their application for higher FAs. Recently, Deng et al. (20) proposed another RF pulse design method, which is based on common slice-selective RF pulses. The RF waveforms are applied with a different time delay on separate transmit coils to create a precompensating z-shim phase along the slice profile. In contrast to the RF pulse design methods stated above, this approach has conceptually short RF pulse durations, robust spatial slice encoding, and low computational complexity. The original approach by Deng et al., which used a custom-built local 4-channel transmit head coil, revealed promising results in recovering the signal but generally left noticeable B1 inhomogeneity effects related to the RF coils' spatial profiles (20). Moreover, the time delay was manually adjusted to match the desired signal recovery level, which hinders its application in clinical practice.

In this work, we propose an extension to the approach by Deng et al. (20) for B1 inhomogeneity mitigation and automatic time-delay determination on the basis of the prevailing B1 and B0 maps. The approach is implemented and integrated in a commercially available 3T MR scanner with two fully integrated whole-body transmit channels. The proposed method was evaluated with experiments in humans using a clinical multi-slice FLASH sequence with a low FA level. In addition, a breath-hold BOLD fMRI study was performed with GRE-echo planar images (EPI) using high FAs of 90 degrees. It is shown that major areas with susceptibility-induced signal loss can be recovered by the proposed automated approach, however, at cost of a signal-to-noise ratio (SNR) penalty.

THEORY

Automatic Time-Delay Determination

The RF-based z-shim approach uses a time-delayed excitation to impose a linear-phase gradient along the slice profile to precompensate the spatially varying through-plane magnetic field gradients. This concept is derived from the basic Fourier formalism, where a shift in the time domain creates a linear phase in the frequency domain. Analogously, in the presence of a slice-selection gradient G_z , a time-shifted RF pulse waveform $b(t)$ results in a phase term along the slice profile $m(z)$. When combining this approach with pTX, a simultaneous z-shim can be achieved by applying $b(t)$ on C separate transmit channels with an individual time-delay τ_c , respectively (20):

$$b_c(t + \tau_c) \leftrightarrow m(z) e^{i\gamma G_z z \tau_c}, \quad [1]$$

where $b_c(t + \tau_c)$ is the RF waveform of the c th transmit channel delayed by τ_c . With increasing time delay of the static RF waveform, the steepness of the phase gradient along the slice profile can be determined, which potentially compensates the prevailing through-plane field gradient G_s . Ideally, the time-delay τ_c at slice-position z_n of N slices is individually matched to cancel out the through-plane phase variation at given TE :

$$G_z \cdot \tau_c(z_n) = G_s(x, y; z_n) \cdot TE \quad [2]$$

However, this approach is not able to fully resolve the required spatially varying phase gradient because the imposed linear phase is globally applied along the slice profile $m(z)$ independent from the in-plane x, y coordinates. On the other hand, the spatially localized sensitivity profiles s_c of the transmit coils combined with individual time-delays τ_c again allow some degree of spatially varying phase distributions. Further, it was shown that the field gradient G_s can be roughly predicted from the fieldmap $B_0(x, y; z_n)$ offset as a first linear approximation (16,21) with $G_s(x, y; z_n) = \alpha \cdot B_0(x, y; z_n)$. Thus, the B0 map being initially acquired for pTX RF pulse optimizations can be reused, and no additional adjustments are required. After adding these assumptions to Eq. [2], the slice- and RF coil-specific time delay can be calculated by means of

$$\tau_c(z_n) = \frac{\alpha \cdot \max[\Delta B_0(x, y; z_n)]_{\mathbf{w}_{s_c}} \cdot TE}{G_z}, \quad [3]$$

where α is the slope describing the functional relation between the frequency offset and through-plane field gradient and $\max[|B_0(x, y; z_n)|]$ the worst-case field offset, and thus most critical field gradient within the coil-specific region of interest (ROI) $\mathbf{W}_{s_c}$. From earlier studies at 3T (16,21), the typical value of α was found between -1.0 and -2.0 $\mu\text{T/m/Hz}$. The coil-specific ROI $\mathbf{W}_{s_c}$ aims to localize the impact of the c th RF coil of the spatial sensitivity profile $s_c(x, y; z_n)$. To match the most suitable transmit coil to a given spatial field gradient distribution $G_s(x, y; z_n)$, a sequential algorithm is used that is capable of handling strongly overlapping coil profiles. First, initial coil-specific ROIs $\tilde{\mathbf{W}}_{s_c}$ are computed by adequate thresholding of the sensitivity profiles. The threshold is defined that all initial coil masks $\tilde{\mathbf{W}}_{s_c}$ completely cover the ROI $\mathbf{W}$ with the smallest possible overlap. Then, the final coil-specific regions of interest $\mathbf{W}_{s_c}$ are subsequently calculated by excluding the previously processed ROIs for $c > 1$: $\mathbf{W}_{s_{c+1}} = \tilde{\mathbf{W}}_{s_{c+1}} - \bigcup_{i=1}^c \tilde{\mathbf{W}}_{s_i}$. Last, the $\mathbf{W}_{s_c}$ correlated with the most critical dephasing gradients to find the proper time delays. To make full use of the RF coils providing high, local spatial sensitivity, the RF coil indices can be ordered according to their average B1 sensitivity prior to the masking and delay calculation process.

Figure 1 illustrates the basic algorithm for an arbitrary number of slices and transmit channels. After the initial estimation of the through-plane B0 field gradient from the prevailing B0 maps, a channel- and slice-specific time delay can be determined by evaluating the impact of the transmit sensitivity profiles. The example was pursued for a local transmit array with rather distinct coil sensitivity profiles and a global 2-channel body transmit array, as used in this study. The latter shows strongly overlapping coil profiles with a large spatial scope such that a single coil is quite sufficient to cover the complete ROI $\mathbf{W}$. Note that solely the worst-case field gradient is considered within the coil-specific ROI to determine the corresponding time delay. The higher the absolute value of the time delay, the higher will be the steepness of the precompensating phase along the slice profile. The sign of the time delay is determined to counteract the direction of the worst-case field gradient in the presence of slice-select gradient G_z . In this example, the second and third transmit channel of the local and the first transmit channel of the global TX array mostly contribute to the precompensation of the signal loss in the frontal orbital cortex, as indicated by $G_s(x, y; z_n)$. Clearly, other more sophisticated approximations are possible. For example, the histogram of the field gradient distribution within $\mathbf{W}_{s_c}$ can be analyzed to avoid extreme values associated with noise and to provide a more robust estimate.

B1 Inhomogeneity Correction

The approach is further extended toward B1 inhomogeneity mitigation on the basis of the spatial domain design approach (22). After defining the optimal time delay per channel and slice, the algorithm aims to calculate proper TX coil-power levels for each individual setting. Please note that prior B1 map normalization and channel-specific ROI processing was solely applied in the context of the time-delay determination. Here, the unprocessed B1 maps information within the complete ROI is used. When discretizing in time and space, an inverse problem must be numerically solved to determine the optimized RF pulse $\mathbf{b}$:

$$\mathbf{b} = \arg_{\mathbf{b}} \min \{ \| \mathbf{A} \mathbf{b} - \mathbf{m} \|_{\mathbf{W}}^2 + R(\mathbf{b}) \}, \quad [4]$$

where $\mathbf{A}$ denotes the overall system matrix, $\mathbf{m}$ the target magnetization pattern, $\mathbf{W}$ the region of interest, and $R(\mathbf{b})$ a regularization term. Essentially, the posed B1 optimization problem is quite similar to the basic RF shimming approach, where a flat target magnetization $\mathbf{m}$ has to be optimized. For this purpose, RF-transmit-channelspecific complex optimization weights w_c are introduced that aim to optimally scale a static RF waveform $\mathbf{p}$. But in contrast to conventional RF shimming, the static RF waveform is not applied simultaneously on all C transmit channels, but the onset of the waveform $\mathbf{p}$ differs from channel to channel due to the introduced time delay τ_c :

$$\mathbf{b}_c(t_j) = \begin{cases} W_c \cdot \mathbf{p}(t_{j-N_{center}} - \tau_c) & N_{center} + \tau_c / \Delta t \leq j < N_{center} + N_p + \tau_c / \Delta t \\ 0 & \text{otherwise} \end{cases} \quad \text{with } j=1 \dots N_t \quad [5]$$

and $\mathbf{k}(t_{N_{center} + N_p / 2}) = 0$,

where $\mathbf{b}_c$ is the final discretized RF waveform of the c th channel with N_t time samples, t the sampling duration, and N_p the number of samples of the static wave-form $\mathbf{p}$. N_{center} denotes the sampling-onset-point from which the center of a symmetric RF waveform aligns with the excitation k-space center. For the optimization process, the basic pulse shape $\mathbf{p}$ is transferred from the solution vector $\mathbf{b}$ to the matrix $\mathbf{A}$ (23,24) such that the actual solution vector is solely composed of the optimization weights $\mathbf{b} = [w_1, \dots, w_C]^T$. In this case, the overall system matrix $\mathbf{A}$ is concatenated by both RF-channel-specific system matrices $\tilde{\mathbf{A}}_c$ and diagonal matrices $\tilde{\mathbf{S}}_c$. The latter hold the information about the RF coil sensitivity s_c at the spatial coordinates $\mathbf{r}$ with N_s samples, respectively.

$$\mathbf{A} = [\tilde{\mathbf{S}}_1 \tilde{\mathbf{A}}_1 \dots \tilde{\mathbf{S}}_C \tilde{\mathbf{A}}_C] \quad \text{with } \tilde{\mathbf{S}}_c = \text{diag} \{s_c(\mathbf{r})\} \quad [6]$$

The i th spatial element a_{ic} of a system matrix $\tilde{\mathbf{A}}_c$ further incorporates information about the time course of the k-space trajectory $\mathbf{k}$ and the evolving off-resonance effects based on the main field inhomogeneities B_0 within the total pulse duration $T_{pulse} = N_t \cdot t$.

$$a_{ic} = \sum_{j=N_{center} + \tau_c / \Delta t}^{N_{center} + N_p + \tau_c / \Delta t} i\gamma m_0 \Delta t e^{i\mathbf{r}_i \mathbf{k}(t_j)} e^{i\gamma \Delta B_0(\mathbf{r}_i)(t_j - T_{pulse})} \mathbf{p}(t_{j-N_{center}} - \tau_c) \quad [7]$$

Note that the time dimension is compressed by a factor N_p by the transfer of the static waveform $\mathbf{p}$ from $\mathbf{b}_c$ to the matrix $\tilde{\mathbf{A}}_c$. Thus, as in RF shimming, the final dimensions of $\mathbf{A}$ are $N_s \times C$, but differ slightly in its composition for $t_c = 0$.

METHODS

RF Pulse Design

The time-delayed RF pulses were calculated in MATLAB 8.0 (MathWorks, Natick, MA) using the magnitude-least-squares approach of (25). Common Hamming-filtered RF sinc pulses were used as static slice-selective RF waveforms $\mathbf{p}$ discretized with $N_p = 200$

samples. Slice- and TX coil-specific RF pulses were optimized using the presented two-step approach. First, optimized time delays were determined matching the coil-specific impact regions to the spatial distribution of the through-plane main field gradients estimated from the B0 map. Then, the set of time delays was considered during the B1 optimization of the TX coil-specific RF waveform weights. To control the degradation of the slice profile (20), high time-bandwidth products were used, and the maximum time delay was limited to 50% of the main sinc lobe duration. Time delays below 10 μ s were discarded due to their low impact. The slice-selection gradient amplitude G_z was set to 19 mT/m. All pulses were regularized to stay within the given RF hardware and SAR limits. No special SAR handling was used in this study beyond the commercially implemented SAR supervision features on the scanner system.

Experiments

All experiments were performed on a 3T MAGNETOM Prisma system equipped with two independent and fully integrated whole-body transmit channels (Siemens AG, Healthcare Sector, Erlangen, Germany). The complex RF coil-sensitivity profiles were measured using a presaturation TurboFLASH sequence (26). A fat-water in-phase B0 map was calculated on the basis of the multi-echo approach similar to (27). A regularized spatial filter was applied to avoid the presence of noisy extreme values (28). The obtained B0 map was incorporated into both the RF pulse calculation (Eq. [7]) and the slice- and RF coil-specific time delays (Eq. [3]). Proof-of-concept human in vivo experiments were pursued for a clinical multi-slice FLASH sequence and breath-hold BOLD fMRI study for three healthy subjects. Breath-hold fMRI is known to create large BOLD contrast over the whole brain; therefore, it is also available to evaluate the method.

FLASH images were acquired with field of view (FOV) 240×240 mm², matrix 256×256 , slices 21, slice thickness 5 mm, TE/TR 20/600 ms, generalized autocalibrating partially parallel acquisition acceleration factor 2, and 25 degrees FA, resulting in a total acquisition time of 1:26 min. Several scenarios were applied using the standard RF excitation in circularly polarized (CP) mode and the presented time-delayed excitation. To highlight the differences to the approach by Deng et al. (20), different time-delay modes were explored: a) the default excitation with a manually optimized global static delay of $\tau_2 = -150$ μ s in CP mode and no B1 inhomogeneity mitigation, b) with the same global static delay including the proposed B1 correction, and c) using both B1 optimization and automatic time-delay determination. In a and b, the global static delay was chosen for the second RF coil, providing generally the higher B1 sensitivity, and was found to offer an appropriate signal recovery across all slices. Four metrics were calculated to quantify signal recovery and image quality. First, the level of signal recovery was assessed similar to (17) by evaluating the percentage of voxels in void regions of the standard RF sinc pulse images recovered by the pTX z-shim pulses. The void regions were defined as areas that fall below 10% of the slice-specific intensity maximum. Background noise was masked out by using the mask **W** gained from the B1 and B0 map adjustments. For the residual brain areas not suffering from signal loss, the sum of squared differences (SSD), standard deviation (STD) of differences, and mean image intensity loss were calculated as a distance to the standard RF sinc acquisitions to quantify the preservation of signal intensity and image quality. In particular,

the SSD metric covers the general loss in SNR and potentially new signal dropouts caused by the time-delayed RF excitation. On the other hand, the STD metric assesses the degree of introduced image inhomogeneity effects, assuming that the RF sinc images provide proper image quality. The mean image intensity loss is stated as a rough estimate for the loss of SNR relative to the RF sinc images. Furthermore, two distinctive slices (i.e., with and without initially strong signal dropouts) were chosen for a more thorough visual analysis. Both were compared in dependence of different RF pulse optimization modes and α values. Difference images were calculated relative to the standard RF excitation case with normalized image intensity.

fMRI breath-hold experiments were pursued consisting of alternating 19.2 s blocks of a breathing/breath-holding paradigm with a total duration of 3:22 min. Sequence parameters were similar as above but with matrix 96×96 , TE/TR 30/3200 ms, and FA of 90 degrees. Here, standard RF excitation was solely compared to the proposed automated time-delayed excitation approach with $\alpha = -2 \mu\text{T/m/Hz}$. fMRI data were processed using the commercially available software (Inline BOLD, Neuro3D, Siemens AG, Healthcare Sector, Erlangen, Germany) available on the scanner. In short, all image time series were motion-corrected and realigned with the first time frame. Regressors were modeled by the convolution of the block-designed breath-hold/normal-breathing paradigm with the canonical hemodynamic response function. Statistics across the 21 slices were calculated for all brain voxels using a general linear model approach (29). The resulting whole-brain t-maps were further evaluated analyzing activation coverage in manually selected ROIs versus the other brain areas at threshold $t > 2.5$.

RESULTS

Image Artifacts with Time-Delayed RF Excitation

To prove the necessity for individually tailored time delays and subsequent B1 inhomogeneity correction, the original method (20) with a static delay is systematically analyzed for two distinct slices across two subjects (Fig. 2). With increasing time delays, a perturbation of the slice profile (i.e., a slight narrowing) becomes observable. At the same moment, signal can be visibly recovered at the prefrontal cortex but at different optimal levels: In subject 1, signal is progressively recovered up to the highest time delay of $-180 \mu\text{s}$. On the other hand, signal is already sufficiently recovered with $-60 \mu\text{s}$ in subject 2 at a similar anatomic position. A further increase of the time delay does not provide any further boost in signal recovery but a deterioration of image quality in the other areas. Similarly, the overall SNR and image homogeneity are generally impaired when applying time-delayed RF pulses to areas without through-plane B0 variations, as is the case in the slices presented in the lower rows of Figure 2. Here, the z-shimming phase gradients lead to undesired signal loss. Consequently, the optimal time delays should be limited to a required minimum to preserve the slice profile and image quality.

FLASH Experiments

A representative subset of FLASH images of volunteer 2 is shown in Figure 3, acquired with different RF excitation modes. From the visual inspection of the data using the standard RF

sinc pulse, typical through-plane susceptibility-induced signal losses can be located in frontal and lower temporal brain areas (Fig. 3a). All of the experiments based on the simultaneous pTX z-shim approach (Figs. 3b–d) can partially recover the signal loss by the application of a time-delayed excitation on two separate whole-body RF excitation coils. First, the images with the original pTX z-shim method (Fig. 3b) were obtained by manually adjusting the time shift to provide the most adequate signal recovery level across all slices. For this purpose, a static time delay of $-150\ \mu\text{s}$ was found to be most suitable for all subjects and was applied to the second RF transmit channel with higher spatial sensitivity. Again, the images indicate that the signal recovery comes along with increased B1 shading effects and a globally reduced SNR. Second, the same time delay was used, but in combination with the proposed B1 optimization approach (Fig. 3c). Here, shading effects were visibly mitigated, but partially at the cost of a lower, spatially dependent signal recovery level (e.g., see lower left image of Figure 3). Finally, the introduction of the automatic time-delay determination offers a slice-dependent time shift (Fig. 3d). For all subjects, the time delays varied from $-40\ \mu\text{s}$ to $-250\ \mu\text{s}$ with $\alpha = -2\ \mu\text{T/m/Hz}$, and they correlate well with the prevailing initial through-plane susceptibility artifact level. The presented algorithm to determine the coil-specific time delays always yielded a single delay associated with the second coil with dominating spatial sensitivity (Fig. 4). In matters of the corresponding spatial profile, the second channel already covers sufficiently the whole FOV. Consequently, the first coil with lower B1 sensitivity showed only minor remaining coil-specific ROIs resulting in negligible time delays. When comparing the proposed method to the B1 shimmed static delay results, no noticeable differences in image quality can be observed at first glance.

However, differences in the signal recovery and image preservation performance become more apparent by studying the proposed quality metrics (Table 1) and difference images (Fig. 5). For the latter, slice 11 (affected by signal loss) and slice 16 (not affected) of Figure 3 were picked for the detailed visual analysis and highlight the shortcomings of the different approaches. Regarding the experiments with a global time delay (Figs. 5a,b), the B1 optimization helps balance the overall signal intensity level of both the recovered and nonaffected areas. This results in an apparent reduction of the signal recovery intensity, but also in less degradation of the average intensity level and image quality [from -55% to -35% , SSD (STD) reduced from 1.00 (1.00) to 0.31 (0.68)]. Still, slice 16 is unnecessarily compromised despite of the absence of initial susceptibility artifacts. The automated time-delay determination tries to overcome this issue by offering a suitable time delay by considering the B0 maps for a through-plane field gradient estimate. The appropriate choice of the α value is important to describe this functional relationship. Figures 5c–e show the results with stepwise decreased α values from -1 to $-3\ \mu\text{T/m/Hz}$. Here, the image quality of slice 16 can be maintained independent from the chosen value. On the other hand, a trade-off between signal recovery level and SNR loss can be observed for slice 11 and in all quality metrics. With higher absolute alpha values, more signal is likely to recover, but involves worse B1 shimming quality and SNR level. Across all experiments, we observed $\alpha = -2\ \mu\text{T/m/Hz}$ as the best compromise between those factors.

fMRI Experiments

In Figure 6, selected slices of the processed breath-hold fMRI experiment are shown for two volunteers. Particularly, the EPI of the first time frame are overlaid with the thresholded BOLD fMRI activation maps for $2.5 < t < 15$ using the standard RF sinc (Figs. 6a,d) and proposed time-shifted spokes excitation (Figs. 6b,d) with FA of 90 degrees. Differences in the spatial activation are further highlighted in Figures 6c,f, where the discrepancy in coverage is illustrated in red (pTX z-shim) and blue (RF sinc) color, respectively. Similar to the FLASH experiments, prominent signal dropouts in the frontal orbital and temporal cortex can be observed in the acquired unprocessed EPI data using the standard RF pulse. Clearly, this results also in a noticeable loss in the corresponding areas of the BOLD activation maps. This through-plane susceptibility-induced signal loss was visibly recovered with the application of the slice-specific time-delayed excitation, but can be partially accompanied with BOLD activation loss in other brain areas. This trade-off can be noticeably recognized in slices of subject 1 suffering from strong susceptibility (e.g., see fourth slice in first row of Figures 6a,b. The average change of the BOLD activation coverage regarding the different RF excitation is further summarized in Table 2 for all subjects. The fMRI sensitivity was compared in manually selected ROIs (green contour lines in Fig. 6) against the other brain areas. Here, the fMRI coverage in through-plane susceptibility-affected areas could be significantly increased by the usage of the simultaneous z-shim approach (up to 70%). At the same time, the BOLD coverage in the remaining brain areas became either slightly reduced or enlarged.

DISCUSSION

In this study, a new method is proposed that reduces through-plane susceptibility-induced signal loss in gradient-echo-based sequences. The method extends the previous work of Deng et al. (20) and is demonstrated to recover more signal, while at the same time the image quality is less compromised. The final method can be used in a fully automated way and is integrated into a pulse sequence on a clinical pTX scanner.

An RF pulse-optimization approach was introduced toward mitigating B1 inhomogeneities in the time-delayed RF excitation. For this purpose, we extended the default RF shimming approach to support RF-transmit channel-specific time-delayed waveforms on the basis of the well-known spatial domain design approach. The FLASH experiments revealed that B1 shading effects could be visibly reduced compared to the original approach, but can also partially reduce the signal recovery level (Figure 5a vs. Figure 5b). This is probably due to the adaptation of the magnitude levels of the arbitrary time-delayed RF sinc waveforms to improve the overall B1 homogeneity. Thus, the impact of those RF coils being responsible for the local signal recovery can be decreased. Furthermore, we observed that the B1 mitigation performance gets generally worse with increasing time delays (see Figure 2 and slice 11 in Figures 5c–e). The more the RF excitation is delayed, the farther its distance becomes relative to the excitation k-space center. Hence, the possibility to adjust the weight of the respective RF coil-sensitivity profile is limited to higher spatial frequencies and no longer can compensate low-frequency B1 shading, which is indicated by the falloff of the second transmitter-sensitivity profile (Figure 4 vs. Figures 5c–e).

The simultaneous z-shim approach was extended by an automated time-delay determination method and aims to adapt the individually required precompensation phase to the susceptibility-provoked signal dephasing. Similar to (16,21), the prevailing B0 maps were used as a rough linear predictor for estimating the through-plane field gradient $G_s(x, y; z_n)$ for each slice. The coil-specific time delay was then determined on the basis of the maximum dephasing gradient value within the corresponding region of interest $\mathbf{W}_{s_c}$. Here, the proposed delay determination is practically limited to a single delay per slice due to the global spatial coverage of the coil-sensitivity profiles (Fig. 4). This certainly restrains the intraslice performance due to the missing adaption to the different, local through-plane main-field variations. The signal-recovery performance is likely to improve with local transmit-coil arrays with more distinct sensitivity profiles. However, at this time-point, there is no commercially available MR scanner system with multiple local-transmit channels for clinical use available at this time. To improve the robustness of the method, the histogram of the gradient values in $\mathbf{W}_{s_c}$ can be analyzed to estimate the major dephasing component being less sensitive to outliers. The tailoring of RF pulse delays to the maximum field gradient alone is critical and relies on sufficient spatial filtering of the B0 map. Still, areas without through-plane B0 inhomogeneity can be affected by the application of phase gradients. Strategies to determine the need for z-shimming can help enhance the adjustment of the time delays. The introduction of a B0 map threshold can be used to indicate whether an effective z-shimming is required for an imaging slice. Especially in dorsal slices with negligible B0 artifacts, this strategy can help avoid unnecessary degradation of the image quality. Furthermore, k-means clustering or advanced thresholding algorithms can be used to improve the definitions of $\mathbf{W}_{s_c}$, which is important for matching the most suited transmit coil with the critical through-plane main-field gradients.

One general drawback of the simultaneous z-shim method remains in the loss of image signal intensity with increasing time delays. This effect was already reported in (20) and is mainly due to the degradation of the slice profile resulting from the superposition of shifted and unshifted RF pulses (Fig. 2a). The slice profile perturbation becomes worse with higher time delays; thus, the SNR loss correlates linearly with the parameter value α (Figs. 5c–e). In contrast to the work by Deng et al., we observed a global loss in SNR because two whole-body RF transmit channels with rather global sensitivity profiles in the brain were used. On the other hand, our approach limits this effect to the required minimum by the adequate choice of slice-individual delays. Furthermore, it offers an efficient signal recovery, short RF pulse durations, and a simple and fast implementation, which is also important for clinical applicability. Based on both clinical multi-slice FLASH and breath-hold BOLD fMRI experiments, the beneficial effects of the proposed methods were demonstrated.

Within the FLASH study, the linear estimate of $G_s(x, y; z_n)$ with a slope of $\alpha = -2 \mu\text{T/m/Hz}$ was found to provide the best compromise between image-quality preservation and local signal-recovery level (Table 2; Figs. 3 and 5). Across all subjects, 47% of the signal could be recovered while limiting the signal intensity loss and inhomogeneity to an adequate level. Compared to the original approach with manual optimization of the time delays, the automated time-delay determination provides approximately the same recovery level, but

with significantly better overall image quality in combination with the B1 inhomogeneity mitigation.

Our proposed simultaneous pTX z-shim approach was applied in a breath-hold BOLD fMRI study with 90-degree RF excitation FA level. Generally, we observed good performance of the small-tip-angle approximation used in the RF pulse optimization for designing the large-tip-angle excitation pulses, which is in line with the early observations of (30). For all subjects, the pTX z-shim approach helped significantly increase the spatial coverage of BOLD activation maps up to 72% within the through-plane susceptibility-affected brain regions compared to the default RF sinc excitation. However, the increased BOLD coverage in signal void areas came along with partial activation loss in the surrounding areas. Particularly, a strong weighting toward signal recovery was recognized by visual inspection in relevant slices of subject 1 (Fig. 6b, lower left slice). Here, the chosen parameter value of $\alpha = -2 \mu\text{T/m/Hz}$ is likely to overestimate the through-plane magnetic field variations. In fact, the study in (21) found $\alpha = -1 \mu\text{T/m/m/Hz}$ to offer the best functional relationship and results in combination with spectral-spatial RF pulses. In this study, we observed insufficient signal recovery when combining this value with the proposed approach (Table 1). As indicated in (13), the used assumption of (16) may generally not be valid to accurately predict the through-plane dephasing for all slices. The relationship was shown to exist for regions above air space, but not necessarily for regions in the same z-plane as air space. Regarding the residual brain areas, the activation coverage could be maintained or even slightly increased (see Table 2). The loss in global SNR, as observed in the FLASH experiments, was not found to noticeably impair the final BOLD contrast. This is in line with previous studies analyzing the fMRI performance with respect to parallel imaging techniques (31,32). It was shown that the BOLD is not necessarily reduced by the loss of image SNR, but is mainly limited by the temporal SNR and physiological noise. However, a more thoroughly study is required to confirm these observations for the pTx z-shim approach.

CONCLUSION

A fully automated simultaneous z-shim approach was presented to recover local signal in GRE-based sequences being impaired by through-plane susceptibility artifacts. The approach includes short RF pulse durations and a robust design for arbitrary FA levels, and it is based on a time-delayed RF excitation on separate RF excitation coils. Thus, a precompensating phase along the slice profile can be imposed to cancel out signal dephasing. We extended this approach to improve both the signal recovery and image quality preservation. The proposed methods take advantage of the prevailing B1 and B0 maps information to optimize the time-delayed RF excitation pulse toward B1 inhomogeneity mitigation and an appropriate slice-individual precompensation phase. Based on these optimized time-delayed RF pulses, clinical multi-slice FLASH and breath-hold BOLD fMRI experiments were pursued on a commercially available clinical pTX scanner system to evaluate their performance. Although the method is generally accompanied with SNR penalties, it was shown to recover 47% of the signal and to improve 57% of the BOLD activation coverage within areas of initially noticeable signal dropouts. Furthermore, the straightforward and fast implementation enables practicability for clinical routine. An

application to systems that have more TX channels with more focused RF-coil sensitivities is of high interest.

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FIG. 1.

Basic automated approach for determining a slice and transmit channel-specific time delay for imposing a simultaneous z-shim on an exemplary 2-channel body or 8-channel local transmit system. Based on the prevailing B0 maps $B_0(x, y; z_n)$ the through-plane field gradient $G_s(x, y; z_n)$ is predicted based on a linear estimate. Further, by considering the spatial sensitivity profiles of the transmit channels, a time-delay $\tau_c(z_n)$ can be sequentially calculated to impose an adequate z-shim within the corresponding region of interest $\mathbf{W}_{sc}$.

FIG. 2.

Imaging artifacts associated with time-delayed RF excitation without B1 inhomogeneity correction. pTX z-shim RF pulses were applied without B1 optimization with static time delays across all slices. Time delays were stepwise increased from 0 to $-180\mu s$ in $60\mu s$ steps (from left to right). (a) Corresponding slice-profiles measured in a phantom. (b) FLASH images of two distinct slices with and without initial susceptibility artifacts. (c) Similar slice positions as in b, but different subject. With increasing time delay, more prominent signal-to-noise ratio penalties, B1 inhomogeneity, and slice perturbations can be observed. The optimal time delay (indicated by the white circle in the upper left corner of the images) providing the best compromise of signal recovery and artifact level varies from slice to slice and subject to subject.

FIG. 3.

Multi-slice FLASH images acquired from one volunteer with flip angle = 25° and different RF excitation modes. **(a)** Standard RF sinc excitation in circularly polarized mode. **(b)** The original simultaneous z-shim approach without B1 optimization and a global static delay of $-150 \mu\text{s}$. **(c)** Same as b, but with B1 inhomogeneity mitigation. **(d)** The final proposed approach including B1 inhomogeneity correction and a slice-specific determination of the time delays. In b–c, the delay was applied solely to the second transmit channel with dominating transmit sensitivity. In d, the time delay associated with the first transmit channel having less B1 sensitivity was practically always resulting to minor values close to $0 \mu\text{s}$ (see Figure 4). Sample areas initially suffering from signal loss caused by through-plane field variation are marked with red arrows. Green arrows indicate areas, which partially show increased B1 shading due to the time-delayed excitation. The two red encircled slices were further analyzed in Figures 4 and 5.

FIG. 4.

Detailed illustration of the proposed time-delay determination of two FLASH images with and without initial through-plane susceptibility artifacts. The signal dropout in slice 11 can be well associated with the corresponding through-plane field gradient maps. The B1 maps reveal that the second radio-frequency channel always provides the superior B1 sensitivity covering quite the whole field of view. After determining the coil-specific ROIs, the remaining mask of the first transmit channel shows only a minor spatial impact. The correlation of the coil-specific ROIs with the estimated through-plane main-field gradient always results in a distinct delay for the second radio-frequency transmit channel and a negligible delay for the first.

FIG. 5.

Detailed analysis of two FLASH images with and without initial through-plane susceptibility artifacts. Difference images were calculated relative to the acquisitions using the standard RF sinc pulse and are normalized to the mean image intensity, respectively. The signal-to-noise ratio loss coming along with the time-delayed excitation is quantitatively captured by the mean intensity loss. Similar to Figure 3, the simultaneous pTX z-shim includes the following modes: (a) The original approach without B1 optimization and a global static delay of $-150 \mu\text{s}$ (mean intensity loss for slice 11/slice 16: $-40.0/-39.0 \%$), (b) with B1 inhomogeneity mitigation ($-21.5/-18.7 \%$), and (c–e) including B1 inhomogeneity correction and slice-specific determination of the time delays with varying α values ($\alpha = 1$: $-9.9/+8.0 \%$; $\alpha = 2$: $-33.9/+5.4 \%$; $\alpha = 3$: $-43.3/+1.8 \%$). pTX, parallel transmit.

FIG. 6.

Breath-hold BOLD functional MRI results of subject 1 (top) and subject 2 (bottom) for t -values > 2.5 . Results were obtained by using the standard RF sinc (**a,d**) and the proposed pTX z-shim approach with B1 mitigation and automatic time-delay determination (**b,e**). The green overlaid contour lines show the manually selected ROIs, where most of through-plane susceptibility artifacts occur. Differences in the spatial BOLD activations are further highlighted in (**c,f**): Red colored areas indicate BOLD activation solely achieved with pTX z-shim and blue areas with RF sinc excitation, respectively. Gray color encodes areas of

common activation. BOLD, blood oxygen-level-dependent; pTX, parallel transmit; RF, radio frequency.

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Table 1**Performance Analysis of Multi-Slice FLASH Images Acquired With pTX**

	pTX Z-Shim Approach	Subject			Average
		1	2	3	
Recovered Signal in [%]	No B1 opt, $\tau = -150\mu\text{s}$	45.8	43.5	43.9	44.4
	B1 opt, $\tau = -150\mu\text{s}$	36.5	42.4	43.1	40.7
	B1 opt, $\tau = \text{auto}$, $\alpha = -1 \mu\text{T/m/Hz}$	41.7	45.8	41.2	42.9
	B1 opt, $\tau = \text{auto}$, $\alpha = -2 \mu\text{T/m/Hz}$	44.7	48.0	47.4	46.7
	B1 opt, $\tau = \text{auto}$, $\alpha = -3 \mu\text{T/m/Hz}$	47.9	50.5	48.9	49.1
SSD	No B1 opt, $\tau = -150\mu\text{s}$	1.00	1.00	1.00	1.00
	B1 opt, $\tau = -150 \mu\text{s}$	0.40	0.31	0.49	0.4
	B1 opt, $\tau = \text{auto}$, $\alpha = -1 \mu\text{T/m/Hz}$	0.24	0.07	0.20	0.17
	B1 opt, $\tau = \text{auto}$, $\alpha = -2 \mu\text{T/m/Hz}$	0.67	0.33	0.69	0.56
	B1 opt, $\tau = \text{auto}$, $\alpha = -3 \mu\text{T/m/Hz}$	0.89	0.59	0.99	0.82
STD of Differences	No B1 opt, $\tau = -150\mu\text{s}$	1.00	1.00	0.98	0.99
	B1 opt, $\tau = -150\mu\text{s}$	0.66	0.68	0.79	0.71
	B1 opt, $\tau = \text{auto}$, $\alpha = -1 \mu\text{T/m/Hz}$	0.64	0.55	0.65	0.61
	B1 opt, $\tau = \text{auto}$, $\alpha = -2 \mu\text{T/m/Hz}$	0.78	0.73	0.88	0.80
	B1 opt, $\tau = \text{auto}$, $\alpha = -3 \mu\text{T/m/Hz}$	0.86	0.82	1.00	0.89
Mean Intensity Loss in [%]	No B1 opt, $\tau = -150\mu\text{s}$	-31.5	-40.7	-32.8	-35
	B1 opt, $\tau = -150\mu\text{s}$	-20.4	-22.9	-20.1	-21.1
	B1 opt, $\tau = \text{auto}$, $\alpha = -1 \mu\text{T/m/Hz}$	-13.2	-15.3	-5.6	-11.4
	B1 opt, $\tau = \text{auto}$, $\alpha = -2 \mu\text{T/m/Hz}$	-28.6	-25.8	-19.0	-24.5
	B1 opt, $\tau = \text{auto}$, $\alpha = -3 \mu\text{T/m/Hz}$	-33.3	-38.5	-27.6	-33.1

Signal recovery performance is measured in percent of recovered voxels < 10% of the normalized maximum intensity in standard RF sinc images. The image quality loss relative to the standard RF excitation in the remaining brain areas is further estimated by the quality metrics SSDs, STDs, and mean image intensity loss. The SSD and STD values were normalized to the highest value per subject, respectively. The mean intensity loss was calculated outside the recovery regions with respect to the RF sinc images, pTX, parallel transmit; RF, radio frequency; SSD, squared differences; STD, standard deviation of differences.

Table 2

Average Change in the Spatial fMRI BOLD Activation Coverage for $t > 2.5$ With the Proposed Simultaneous pTX Z-Shim Approach

	Subject			Average
	1	2	3	
Recovery region (ROI)	+72.5 %	+42.3%	+58.2%	+57.6%
Outside recovery region (ROI)	-5.9 %	+9.6%	+8.8%	+4.2%

Values are stated in percent relative to the results of the standard RF sinc excitation pulse for the manually selected ROI and the other remaining brain areas (ROI), respectively. fMRI, functional MRI; BOLD, blood-oxygen-level-dependent, pTX, parallel transmit; ROI, region of interest.