Full Paper

Fast online-customized (FOCUS) parallel transmission pulses: A combination of universal pulses and individual optimization

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Purpose: To mitigate spatial flip angle (FA) variations under strict specific absorption rate (SAR) constraints for ultra-high field MRI using a combination of universal parallel transmit (pTx) pulses and fast subject-specific optimization.

Methods: Data sets consisting of B₀, B₁⁺ maps, and virtual observation point (VOP) data were acquired from 72 subjects (study groups of 48/12 healthy Europeans/Asians and 12 Europeans with pathological or incidental findings) using an 8Tx/32Rx head coil on a 7T whole-body MR system. Combined optimization values (COV) were defined as combination of spiral-nonselective (SPINS) trajectory parameters and an energy regularization weight. A set of COV was optimized universally by simulating the individual RF pulse optimizations of 12 training data sets (healthy Europeans). Subsequently, corresponding universal pulses (UPs) were calculated. Using COV and UPs, individually optimized pulses (IOPs) were calculated during the sequence preparation phase (maximum 15 s). Two different UPs and IOPs were evaluated by calculating their normalized root-mean-square error (NRMSE) of the FA and SAR in simulations of all data sets. Seven additional subjects were examined using an MPRAGE sequence that uses the designed pTx excitation pulses and a conventional adiabatic inversion.

Results: All pTx pulses resulted in decreased mean NRMSE compared to a circularly polarized (CP) pulse (CP = ~28%, UPs = ~17%, and IOPs = ~12%). UPs and IOPs improved homogeneity for all subjects. Differences in NRMSE between study groups were much lower than differences between different pulse types.

Conclusion: UPs can be used to generate fast online-customized (FOCUS) pulses gaining lower NRMSE and/or lower SAR values.
1 | INTRODUCTION

Ultra-high field (UHF) MRI (B₀ ≥ 7T) yields improved SNR¹ and, often, also improved image contrast. Therefore, it provides unprecedented anatomic detail in neuroimaging.² However, there are also substantial challenges that need to be overcome to translate the full potential of UHF MRI in the clinic.³ In particular, the spatial variation in the transmit magnetic field (B₁⁺) results in spatially varying flip angles (FA) leading to undesirable variations in signal intensity and image contrast. In addition, spatial variation of the local specific absorption rate (SAR) is becoming more dominant at higher field strengths.

To overcome these limitations, the concept of parallel transmission (pTx) has been applied successfully.⁴ PTx makes use of multi-channel transmit (Tx) RF coils that are driven by individual, independent RF pulses and provide an effective means of optimizing the FA distribution while minimizing local and global SAR. Static pTx,⁴ often denoted as B₁⁺ shimming, uses a single RF shape for all Tx channels with individual channel amplitudes and phases designed to optimize B₁⁺. This technique has made a strong impact on mitigating local and global B₁⁺ field inhomogeneities in the human head,⁵,⁶ aorta,⁷ prostate,⁸ and whole body,⁹ yet, has limited effect in highly inhomogeneous areas of the B₁⁺ field. Dynamic pTx typically consists of both a “transmit k-space” trajectory and corresponding individual RF shapes for each Tx channel. During the transmission of a dynamic pTx RF pulse, the resulting B₁⁺ field is usually spatially inhomogeneous at a given time point and temporally changing, but the aim is to achieve a homogeneous FA distribution at the end of the pulse.¹⁰⁻¹² Therefore, dynamic pTx pulses provide more degrees of freedom and, thus, higher potential to produce uniform FA patterns than static pTx pulses. Several techniques to design dynamic pTx pulses have been proposed including slice/slab selective,¹³,¹⁴ tailored 3D volume excitation,¹⁵,¹⁶ and nonselective¹⁷,¹⁸ pulses.

Applying a subject-specific pTx pulse first requires the mapping of B₁⁺ and B₀ within the region of interest and, then, the calculation of RF pulse shapes and potentially gradient trajectories. Presently, the mapping and calculation step hinder a wider application of such pulses, because both steps may require ~15 min total.¹⁹,²⁰ Therefore, they are often regarded as not suitable for clinical routine. A recent major step toward clinically applicable pTx pulses is the concept of “universal pulses” (UPs) by Gras et al.,²¹ which do not require any online calibration. Using B₁⁺ and B₀ maps of multiple training subjects, pTx pulses have been designed that produced much more homogeneous FA distributions than circularly polarized (CP) pulses, which are considered to be equivalent to single Tx, in subjects not included in the training set. Non-selective UPs have been presented for both T₁- and T₂-weighted imaging²¹⁻²³ and slice-selective UPs have been created for T₁ weighted imaging.²⁴ Furthermore, the technique has also been shown to be applicable for various pTx head coils.²⁴ Nevertheless, although this technique allows substantially improved excitation homogeneity without the need for online pulse calculation, Gras et al. also demonstrated that subject-specific pTx pulses (i.e., pulses that are not designed based on the training set but on B₁⁺ and B₀ maps of the actual subject) achieve even higher spatial signal homogeneity.²¹

In this work, we present a method that combines both UPs and a fast subject-specific online calibration requiring <1 min additional sequence preparation time. This short preparation time paves the way for the application of individually optimized pulses (IOPs) in a clinical routine. We call these IOP pulses as fast online-customized (FOCUS) pTx pulses. In the first step, parameters to define the k-space trajectories, energy regularization weights, universal pulse shapes, and fitting curves for maximum local specific energy dose (SED) with respect to further adjustable regularization weights were optimized offline using 12 training data sets. Based on these parameters, subject-specific RF pulse calculation can be performed online during the sequence preparation phase after rapid B₀ and B₁⁺ mapping. Stable improvements of the pulses’ performance regarding FA homogeneity and maximum local SED exposure (often considered to be the most restrictive constraint)²⁵ are shown in Bloch simulations of an additional 60 subjects (36 healthy Europeans, 12 Asians, and 12 Europeans with pathological or incidental findings).

Unlike the original UPs,²¹ the basic trajectory used here is based on a “SPINS” (spiral-nonselective) trajectory¹⁸ with several additional parameters to create more degrees of freedom. These FOCUS pTx pulses were designed for a low FA excitation in a 3D MPRAGE sequence. We demonstrate improved excitation behavior of IOPs as compared to UPs and a reliable performance within simulation across all study subjects. MPRAGE images using CP pulses, UPs, and IOPs were acquired from 7 additional subjects to prove consistency.

2 | METHODS

2.1 | Measurement system and data acquisition

All measurements were conducted on a 7T whole-body MR system (MAGNETOM Terra, Siemens Healthcare GmbH, Erlangen, Germany) with an 8Tx/32Rx head coil
B₀ mapping was performed with a sagittally oriented gradient echo sequence (TE₁ = 2.39 ms, TE₂ = 4.59 ms, TE₃ = 7.09 ms; resolution = 4 × 4 × 6 mm³; 28 slices; no slice gaps; T₁ = 12 s). B₁⁺ mapping was performed using an interferometric transversely oriented magnetization prepared saturation recovery turbo flash sequence (TE = 1.63 ms, TR = 3.76 s, resolution = 4 × 4 × 5 mm³, 30 slices, 2.5 mm slice gapping, T₂ = 40 s), and involved 8 GRE images, where always 1 transmit channel was left out. Absolute single channel B₁⁺ maps were calculated using these 8 GRE images and a reference map (all channels combined) derived from those. A mask was generated based on the B₁ map of the CP mode, excluding voxels with intensities in the lower 10th percentile. Additionally, voxels, whose phases in prepared and unprepared GRE images differed by more than 80° were regarded as erroneous measurement points and consequently also excluded. No brain extraction was performed, and lower and peripheral regions (upper cervical spine, skull, nose, and jaw) were kept because those regions might also be relevant for diagnostics.

In total, 72 subjects were investigated in simulations (41 female, 31 male, age: 30.3 ± 12.9 years, ranging from 19-71 years). The following timing parameters were applied: TR = 3 s, TI = 1.1 s, acquisition time = 7 min 11 s, minimum TE (either TE₁ = 2.92 ms using CP rectangle pulses of 100 µs duration or TE = 3.37 ms using UPs or FOCUS pulses of 1 ms duration), matrix size: 384 × 336 × 256, FOV = 250 × 218.75 × 167 mm³, 0.65 mm isotropic resolution, BW = 250 Hz/Px, GRAPPA acceleration factor 3, and echo spacing (ES) = 6.9 ms using CP rectangle pulses and 7.8 ms using pTx pulses. The transmitter voltage was adjusted by a vendor provided routine consisting of a fast saturation recovery turbo FLASH B₁⁺ mapping sequence that measures B₁⁺ in 3 transversal slices (1 located in the isocenter and 2 located ±4 cm off-center). The transmitter voltage is then set such that the upper 20th percentile of F reaches at least the nominal value. All B₁⁺ and B₀ maps were interpolated into a 4 × 4 × 6 mm³ matrix in sagittal orientation covering the FOV of the sequence.

In total, 72 subjects were investigated in simulations (41 female, 31 male, age: 30.3 ± 12.9 years, ranging from 19-71 years). The following groups were created: the training group (n = 12, 25 ± 2.8, from 21-29 years), the validation groups (n = 60 in total), which included other healthy Europeans (n = 28, 30.0 ± 11.1 years, ranging from 21-63 years), Europeans with certain anomalies or pathological conditions (n = 12, 50 ± 15.3, ranging from 28-74 years), and healthy Asians (n = 12, 26 ± 2.8 years, ranging from 21-31 years). The anomalies include arachnoid cysts (n = 1), multiple sclerosis lesions (n = 3), cavernoma (n = 1), meningioma (n = 3), aneurysm of a middle cerebral artery (n = 2), adenoma (n = 1), and parenchyma defects from traumatic brain injury (n = 1).

In addition to Bloch simulations that were performed on each subject and for all pulses, 7 additional European subjects were examined experimentally using the MPRAGE sequence in combination with FOCUS pulses. Six of these patients were healthy, and one was a brain surgery patient, who was considered to have the most apparent anomalies among all. The study was approved by the local Ethical Review Board, and all subjects provided informed consent before the scan.

2.2 k-Space trajectory parametrization and individual RF pulse optimization

The transmit k-space trajectory used was based on a 3D SPINS. SPINS pulses provide more pulse samples, and therefore, more degrees of freedom for RF pulse optimization as compared to the kₜ point trajectory. To increase the degrees of freedom for the trajectory, several new parameters were added. In spherical coordinates (kₜ, kₚ, and kᵦ denoting the radial, polar, and azimuthal components), the modified trajectory is defined by:

\[ k_{0}(t) = \theta_0 + \omega_0 t + v_0 t^2 \]
\[ k_{\phi}(t) = \phi_0 + \omega_\phi t + v_\phi t^2 \]
\[ k_{z}(t) = \frac{k_{max}}{1 + e^{\alpha (\gamma - \beta)}} + e^{\gamma (\gamma - \delta)} \]

\[ k_{max} \] describes the maximum distance from the k-space center, and \( \alpha \) and \( \beta \) (0 < \( \beta < 1 \)) are parameters describing the evolution of \( k_z \) based on the original SPINS trajectory, and a second order term with parameters \( \gamma \) and \( \delta \) (0 < \( \delta < 1 \)) was added. \( \theta_0 \), \( \phi_0 \), \( \omega_\theta \), \( \omega_\phi \), \( v_\theta \), and \( v_\phi \) indicate the initial angles, angular velocities and acceleration in polar and azimuthal direction, respectively. \( \theta_0 \) and \( \phi_0 \) were added to introduce more degrees of freedom and to avoid high slew rates at the beginning of the trajectory for the x \((k_x = k_{\phi} \cdot \sin k_{\theta} \cos k_{\phi})\) and z \((k_z = k_{\phi} \cdot \cos k_{\theta})\) gradient coils. In addition, the acceleration terms \( v_\theta \) and \( v_\phi \) as well as \( \gamma \) and \( \delta \) were added to provide second order terms. After transforming the spherical coordinates into Cartesian coordinates, the z coordinate of the trajectory was scaled with a constant factor \( k_z(t) = c_k k_z(t) \). This was done to introduce yet 1 more additional degree of freedom for the trajectory in z direction (head–feet), because all transmit elements are located at the same z-position (head–foot direction).
To optimize the individual RF pulse profile, the following problem was solved, which aimed to minimize the squared deviation of the FA across the head volume from a target FA using the variable exchange method\textsuperscript{27,28},

$$\min_{\mathbf{b}} \| \mathbf{A} \mathbf{b} - \mathbf{\alpha}_p \|_2^2 + \lambda \| \mathbf{b} \|_2^2,$$  \hspace{1cm} (2)

where $\mathbf{\alpha}_p = 7^\circ$ denotes the target FA, and $\mathbf{A} \in \mathbb{C}^{N_v \times (N_c - N_h)}$ denotes the spins’ dynamics matrix for all $N_v$ head voxels of a single subject with $N_c = 8$ Tx channels, each having $N_S$ pulse shape samples (detailed description in Hoyos-Idrobo et al\textsuperscript{29}). The matrix $\mathbf{B}$ is defined by the $\mathbf{B}_{ij}^m$ maps with individually generated masks described above (see Measurement system and data acquisition), as well as the k-space trajectory. $\mathbf{b} \in \mathbb{C}^{N_c - N_h}$ describes the concatenated complex waveforms of all Tx channels and $\lambda$ is the energy regularization weight. A fixed pulse duration of $T = 1$ ms at a sampling rate of 100 kHz was used, resulting in $N_S = 100$ pulse shape samples. To enable fast online-customization, we used a single regularization weight $\lambda$ to meet both local and global SAR constraints even though a better trade-off between local and global SAR might be achieved using a combination of $k_T$ points and simultaneously constraining local SAR and power as shown by Guérin et al.\textsuperscript{30}

### 2.3 Universal pulses and energy regularization weights

For the following optimizations, $\mathbf{B}_1^+ \text{ and } B_0$ maps from $N_p = 12$ healthy European training subjects (6 female, 6 male, age = 25 ± 2.8 years, ranging from 22-29 years), as well as virtual observation point (VOP) data\textsuperscript{31} were used as input data. Offline calculations described below (see Equations 6-8) were performed on a desktop computer (Intel Core i7-6700K CPU; 2.7 GHz; 4 Kernels; 64 GB of RAM) using MATLAB 2017b (The MathWorks, Natick, MA) with the parallel computing toolbox.

In a first step, the SPINS trajectory parameters and the (initial) energy regularization weight $\lambda_0$ (see Equation 2) were optimized simultaneously. Consequently, their values were defined as combined optimization values (COV):

$$\mathbf{COV} = \left[ k_{\text{max}}, \mathbf{\alpha}, \mathbf{\beta}, \gamma, \mathbf{\delta}, 0_0, 0_v, 0_0, 0_0, 0_0, 0_0, 0_0, 0_0, 0_0, c_2, \lambda_0 \right].$$ \hspace{1cm} (3)

During the optimization of the COV (Equation 6), for each set of COV, individual RF pulse calculation (Equation 2) was performed. For evaluation, the normalized root mean square error (NRMSE) of the FA and the pulse’s maximum local SED were minimized for each individual pulse. For a single subject $p$, the NRMSE was calculated as

$$\text{NRMSE}_p = \frac{1}{N_v \alpha_m \sum_{v=1}^{N_v} (\alpha(v) - \alpha_m)^2},$$\hspace{1cm} (4)

using a Bloch simulation where $\alpha(v)$ denotes the simulated FA in voxel $v$ among $N_v$ voxels in total, and the overall mean FA is denoted by $\alpha_m$. The SED was chosen as a measure of energy absorbed per mass of tissue in [J/kg], which is proportional to the pulse’s contribution to the maximum local SAR in [W/kg]. To specify the individual maximum local SAR values, the $N_{\text{VOP}} = 8$ VOPs derived from 3 different models (Duke, Ella, and Hugo)\textsuperscript{32,33} simulating SAR in each 10-g volume in the subject’s head were used. The VOPs and head (including shoulders) models were provided by the manufacturer of the MR system (Siemens Healthcare GmbH, Erlangen, Germany) and are generally available on the new type of 7T system. The maximum local SAR value for a single subject $p$ from these different VOPs was calculated via the following formula:

$$\text{SED}_p = \arg \max_{j \in \{1...N_{\text{VOP}}\}} \mathbf{R} \left( \| \mathbf{G}_p \mathbf{G}_p^* \otimes \mathbf{B}^j \mathbf{B} \circ \mathbf{VOP} \|_1 \right).$$ \hspace{1cm} (5)

The matrix $\mathbf{B} \in \mathbb{C}^{N_c \times N_c}$ describes the complex waveforms of all channels, containing the same information as the vector $\mathbf{b}$, and the matrix $\mathbf{VOP} \in \mathbb{C}^{N_{\text{VOP}} \times N_c}$ denotes the SAR estimation matrix of VOP $j$. $\mathbf{G}_p \in \mathbb{R}^{1 \times N_c}$ denotes the subject-specific gains on each transmit channel (difference between measured and nominal voltage supplied by the RF power amplifier, measured automatically during each examination), $\mathbf{R}(\ldots)$ denotes the real part of a number, $\otimes$ denotes the Hadamard product, and $\| \ldots \|_1$ denotes the sum norm of a matrix. All other VOP types (e.g., global SAR and hardware protection) were neglected because they were found to be much less constrictive.

To obtain the most suitable COV, the following cost function was minimized:

$$\min_{\mathbf{COV}} \sum_{p=1}^{N_p} \left[ w_{\text{Hom}} \exp \left( \text{NRMSE}_p(\mathbf{COV}) \right) + w_{\text{SED}} \exp \left( \text{SED}_p(\mathbf{COV}) \right) \right].$$ \hspace{1cm} (6)

To evaluate a single cost value on a set of COV, individual RF pulse shape ($\mathbf{b}$) optimization (Equation 2) was performed for all $N_p$ training data sets based on the COV and starting with CP mode and rectangular pulse shapes. $\text{NRMSE}_p$ and $\text{SED}_p$ were derived from the individually optimized pulse for subject $p$ and applied on subject $p$. For this optimization, we used MATLAB’s global search algorithm (“GlobalSearch”, see Ugray et al\textsuperscript{34} for more details). For the input variables (COV), fixed lower and upper bounds (Supporting Information Table S1) and a boundary condition to guarantee feasibility with respect to a maximum slew rate of 200 mT/m/ms were established. A single function evaluation took 234 s and a time constraint of 1 week for the optimization was set. This approach enables a tradeoff, which is adjustable with the ratio of 2 different relative weights $w_{\text{Hom}}$ and $w_{\text{SED}}$ to achieve both good $B_1^+$ homogeneity and low
SAR, respectively. The exponential behavior of each single data set’s contribution to the cost was chosen to strongly penalize single patients’ high NRMSE and SED values aiming to generate pTx pulses, which work universally.

In a second step, when an optimized set COVopt has been found they were used to generate corresponding universal pulse shapes \( b_0 \). This was done by solving the following problem using the same interior-point-based global search method with starting points generated in the same way as described above:

\[
\min_{b_{0, COVopt}} \sum_{p=1}^{N_p} \left[ w_{\text{Hom}} \exp\left( \text{NRMSE}_p \left( b_{0, COVopt} \right) \right) + w_{\text{SED}} \exp\left( \text{SED}_p \left( b_{0, COVopt} \right) \right) \right].
\]  

(7)

Here, a single cost value was calculated via all training subjects’ NRMSE and SED for a specific universal pulse shape \( b_{0, COVopt} \). A single function evaluation took 168 s on the same desktop computer and a time constraint of 1 week for the optimization was set. The magnitude of each RF sample in \( b_{0, COVopt} \) was limited to 150 V (option “nonlcon” in MATLAB’s function “createOptimProblem”).

### 2.4 Individually optimized pulses

Based on the so created UPs (trajectory derived from COVopt and corresponding universal RF shape \( b_{0, COVopt} \)), individual RF shape optimization was performed (Equation 2) for each training subject using varying energy regularization weights \( \lambda \). For each regularization weight, each training subject’s individual maximum SED value of the subject’s respective pulse was calculated. Thereby \( N_p \) different SED data points were generated for each value of \( \lambda \). The following function was used to fit these points:

\[
\text{SED}_{\text{COVopt}, \lambda}(\lambda) = a\lambda^b + c.
\]  

(8)

The parameters \( a, b, \) and \( c \) are specified by minimizing the RMSD of the data points from the fitting curve. This exponential fitting curve was found empirically and outperformed polynomial, linear, sum of sine, and Fourier fitting curves with respect to RMSE. Based on the 10-s limit for local SAR, because it was found to be the most restrictive constraint (20 W/kg; according to IEC guideline 60601-2-33), the given pulse sequence parameters (FA, TR, number of pTx pulses per TR) and the SED of the fixed adiabatic hyperbolic secant inversion pulse (nominal voltage 350 V, pulse duration 12.8 ms, SED = 34.03 ± 1.25 J/kg), a limit for the pTx excitation pulse’s SED was also calculated (SED\(_{\text{ub}}\): 122 ± 5.9 mJ/kg). This SED(\( \lambda \)) curve enables quick adjustment of \( \lambda \) online to reach SED values below the pulse’s SED limit (see section below).

Based on a set of COV (Equation 6), \( b_0 \) (Equation 7) and SED(\( \lambda \))(Equation 8) data that were calculated offline, individual pulse optimization was performed online during the scan on the scanner control computer (Intel Xeon E5-1620 CPU; 3.5 GHz; 4 Kernels; 32 GB of RAM) using MATLAB Runtime 8.0. RF pulse optimization could be performed within 15 s by solving Equation 2. In the case that SAR exceeded the allowed limit, \( \lambda \) was adjusted using an online SED regularization algorithm described in the Supporting Information Pseudocode S1.

All online optimization algorithms were embedded into the scanner’s software system to enable short online computation times and to facilitate future clinical applicability.

The total preparation time of the pulse sequence was 90 s. During this period, standard preparation (transmitter adjustment, frequency adjustment, adjustments of the directional couplers [DICO], \( B_0 \) mapping, \( B_1^+ \) mapping, and pTx pulse design were performed automatically. Default third order \( B_0 \) shim currents using the entire brain as target ROI were applied to all measurements. An overview of the whole process to design FOCUS pulses is shown in Figure 1.

For this study, various sets of \( \text{COV}, b_0, \) and \( \text{SED}(\lambda) \) data were calculated with different relative weightings. The most relevant combinations were evaluated and further examined in this paper, being [\( w_{\text{Hom}}, w_{\text{SED}} \] = [1,1] and [5,1]. These 2 combinations have been chosen among others, because they reflect 2 different tradeoffs: 1 combination that puts weight on low NRMSE values [5,1] whereas still providing not excessively high SED values and 1 set [1,1] that puts weight on low SED while still providing acceptable homogeneity. Consequently, the resulting UPs or IOPs are denoted by UP\(_{11}\), UP\(_{51}\), IOP\(_{11}\), and IOP\(_{51}\). Their respective trajectories are shown in Figure 2 and the optimized COV of both pulses are shown in Supporting Information Table S1. Evaluations of other combinations of [\( w_{\text{Hom}}, w_{\text{SED}} \)] can be found in Supporting Information Figure S1.

### 3 RESULTS

The simulated NRMSE and SED values of all 72 subjects for all pTx pulses and a fixed CP rectangular shaped pulse are reported in Figure 3. The NRMSE values of 28.2 ± 2.4% (mean ± SD) by the fixed CP rectangle pulse could be reduced to 18.3 ± 2.2% by UP\(_{11}\) and 16.2 ± 1.9% by UP\(_{51}\). Individual optimization could substantially further decrease NRMSE to 13.4 ± 0.8% (IOP\(_{11}\)) and 10.8 ± 0.7% (IOP\(_{51}\)). Yet, pTx pulses with [\( w_{\text{Hom}}, w_{\text{SED}} \] = [1,1]) also slightly increased the maximum local SED exposure from 27.49 ± 1.0 mJ/kg (CP) to 30.0 ± 0.96 mJ/kg (UP\(_{11}\)) and 35.9 ± 5.2 mJ/kg (IOP\(_{11}\)). Higher weighting of homogeneity leading to lower NRMSE values, also causes higher amounts of SED, such as 83.5 ± 2.7 mJ/kg (UP\(_{51}\)) and 111.5 ± 19.1 mJ/kg (IOP\(_{51}\)). It has to be noted that the CP rectangle pulse has a shorter duration of 0.1 ms than the 1 ms pTx pulses and a lower mean FA (CP = 4.0°, UP\(_{11}\) = 6.5°, UP\(_{51}\) = 6.6°, IOP\(_{11}\) = 6.8°, and IOP\(_{51}\) = 6.9°). IOP\(_{11}\)
outperformed UP$_{51}$ in both NRMSE (mean decrease by 17%) and SED (mean decrease by 57%). Although all pTx pulses were only trained with healthy European subjects, differences regarding homogeneity between study groups were marginal for all pulses. For each pTx pulse, mean NRMSE values were highest for the “European, anomalies”. 

FIGURE 1  Schematic description of the pTx pulse design workflow. The offline optimization process uses previously measured data sets and predefines the transmit k-space trajectory, initial pulse shapes, SED dependence of the energy regularization parameter, and its initial value. Subject-specific pulse shapes were calculated online using the $B_1^*$ and $B_0$ maps were acquired during preparation time. Therefore, the amount of additional time required for online optimization is 67 s.

FIGURE 2  k-Space trajectories of both optimized UP and corresponding FOCUS pulses UP$_{11}$/IOP$_{11}$ and UP$_{51}$/IOP$_{51}$. The individual sampling points are denoted as black points, with the center part of k-space sampled more densely than the peripheral regions.
group, and lowest for the “European, training” group, yet, those differences were generally minor (Figure 3).

Pulse energies for all subject groups and for all pulse types are displayed in Supporting Information Figure S2. In addition, different IOPs were designed based on various training groups and evaluated on all data sets (Supporting Information Figure S3). Compared to the choice of \(\mathbf{w}_{\text{Hom}}\), \(\mathbf{w}_{\text{SED}}\), the choice of the training group had only minor influences on the results regarding pulse performance. We also optimized pulses using various numbers of training subjects (Supporting Information Figure S4), yet, both increasing the number of subjects to 24 (2 weeks optimization time) and using only single subjects (2 days optimization time) did not improve the results. Furthermore, IOPs trained with single subjects did not show lower cost values for the respective subject it was trained on.

UPs based on SPINS trajectories markedly improved NRMSE compared to CP-mode and they also proved useful as starting point (\(\mathbf{b}_0\)) for fast online optimization of subject-specific pulse shapes (Equation 1), as Figure 4 shows. RF pulse optimization was stopped when the cost function decreased by \(<0.1\%\) of its previous value. When initialized with universal RF shapes, IOP11 reached up to 5% better cost values (average = 1%) in 0.3 s less time on average, and IOP51 reached up to 41% better cost values (average = 8%) in 2.3 s less time on average on our PC.

Figure 5 shows the performance of IOPs with online SED regularization for different limits. Lower NRMSE values are reached by those pulses whose initial SED distributions are closer to the given SED limits (low mean_\(\Delta\)SED values). Doubling the upper SED limit from 100 mJ/kg to 200 mJ/kg resulted in a relatively low improvement of NRMSE (IOP11 = 0.26%, IOP51 = 0.49%), whereas changing from 30 mJ/kg to 100 mJ/kg resulted in a higher decrease of NRMSE (IOP11 = 2.58%, IOP51 = 9.48%). Because both IOPs generate more SED than their corresponding UPs, Supporting Information Figure S5 shows both IOPs regularized with SED-bounds equal to the maximum and minimum SED values of their corresponding UPs to evaluate the benefit of individual RF shape optimization. When generating the SED values in the same bounds, both IOPs still show lower NRMSE values than the corresponding UPs.

Figure 6 shows MPRAGE images and corresponding simulated FA maps of a single subject (m26) using either CP pulses, UP11, UP51, IOP11, or IOP51 without SED regularization as excitation pulses. Both UPs gained higher signal homogeneity and intensity especially in the cerebellum and in the lower head regions compared to the CP mode. IOPs further improved signal homogeneity throughout the entire

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**FIGURE 3** NRMSE and maximum SED values of different pulse types for all 72 subjects grouped in Europeans used for optimization (12), other Europeans (36), Asians (12), and Europeans with anomalies (12). The asterisk denotes the mean value, the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively and the red crosses denote outliers. UP11 and IOP11 were able to decrease the NRMSE compared to the CP pulse, UP11 requires a 9% higher, IOP11 requires a 31% higher amount of SED. Using the same pairs of \(\mathbf{w}_{\text{Hom}}\) and \(\mathbf{w}_{\text{SED}}\), IOPs further improved the homogeneity at the expense of higher SED exposure. Yet, IOP11 reaches both lower NRMSE and SED than UP51. Differences between study groups were generally lower than differences between pulse types.
**FIGURE 4** Online RF shaping cost function decrease for both IOPs as a function of the iteration i using either UP shapes or CP mode as starting points ($b_0$), indicated with “UPinit” (green) and “CPinit” (blue), respectively. The 2 upper plots show the mean curves among the validation data ($n = 60$). The lower plots show the corresponding individual curves. The optimization stopped, when the cost improves $<0.1\%$ of its own value in 1 iteration. When initialized with universal RF shapes, IOP$_{11}$ reached up to $5\%$ better cost values in 18 iterations instead of 71 iterations (when initialized with CP mode and rectangular pulse shapes) on average; IOP$_{51}$ has reached up to $41\%$ better cost values in 13 iterations instead of 35 iterations on average.

**FIGURE 5** NRMSE and maximum SED values of IOP$_{11}$ and IOP$_{51}$ regularized with different upper and lower bounds ($\text{SED}_{ub}$, $\text{SED}_{lb}$, red dashed lines) using all validation data. Their mean values are indicated with red boxes, the corresponding mean values of IOPs without regularization are shown as green boxes. For the same SED limits, IOPs with smaller mean $\Delta$SED (distance between red and green boxes) achieve lower NRMSE. Lowering $\text{SED}_{lb}$ although keeping $\text{SED}_{ub}$ ($[\text{SED}_{ub}, \text{SED}_{lb}] = [100, 80]$ and [100, 50]) can result in lower SED and higher corresponding NRMSE values for single subjects (see outliers). Therefore, by adjusting $\text{SED}_{lb}$, one can give greater weight to achieving low NRMSE values compared to the importance of achieving lower SED values.
head. Yet, differences between UPs with different weights, as well as between IOPs with different weights are smaller than between UPs and IOPs. For better comparison of CP and FOCUS pulses, MPRAGE images and corresponding simulated FA maps of another single subject (m29) including a CP pulse having higher voltage to achieve a mean FA of 7° next to UP11 and IOP11 are shown in Figure 7. Here, both UP11 and IOP11 show a more homogeneous FA distribution than the CP pulses. The differences between both UPs and their corresponding IOPs are shown in Figure 8 on 4 exemplary subjects. FA variations produced by UP11 can be minor (subject 1), but also apparent (subject 2), whereby IOP11 reliably improves homogeneity, especially in the latter case. UP51 also shows inhomogeneities (subjects 3 and 4), which are improved by switching to IOP51. Figure 9 shows several MPRAGE images and FA maps of the subject considered to have the most apparent anomalies. UP11 still yields acceptable signal and FA homogeneity in most brain regions, yet, in some regions, certain anatomical structures (especially parts of the cerebellum) are not visible. IOP11 shows improvement in the FA homogeneity,
having no such regions with severe signal loss. In none of the acquired MPRAGE images artifacts have been identified, which could be attributed to the adiabatic pulse.

4 | DISCUSSION

In this work, we combined the concept of UPs\textsuperscript{21} and individual pulse optimization to achieve a fast online calculation for subject-specific or FOCUS pTx pulses. The individual pulse optimization under strict SAR constraints required only ~67 s of additional sequence preparation time (B\textsuperscript{1+} mapping and pulse calculation). Therefore, the applicability of these pulses is clinically feasible. UPs reliably reduced FA-NRMSE compared to CP mode requiring no additional preparation time, and the proposed FOCUS pulses further improved pulse performances by further decreasing NRMSE. Because only marginal differences in image quality and contrast between MPRAGE images obtained with IOP\textsubscript{11} and IOP\textsubscript{51} were observed, we consider IOP\textsubscript{11} better suited for clinical routine because of its decreased SED exposure. In addition, an online SED-regularization method was implemented that allowed us to explore the full set of available SEDs tailored to subject and sequence parameters. By setting a lower bound for SED, we can enforce lower NRMSE values for subjects, whose SED(λ)-function is strongly differing from the SED fitting curve. By using FOCUS excitation pulses with controlled SED exposure, it was possible to apply a standard adiabatic inversion pulse, which worked sufficiently well.

In contrast to the original UPs, as proposed by Gras et al,\textsuperscript{21} which are based on k\textsubscript{T} points,\textsuperscript{17} we used a SPINS trajectory\textsuperscript{18} that allows distribution the RF energy along an excitation k-space trajectory that includes a larger number of k-space points as than the k\textsubscript{T} points trajectory. This trajectory was originally designed for 3T MRI, and in this work, we introduced several additional parameters that characterize the trajectory to increase the degrees of freedom. Furthermore, the concept of FOCUS pulses can also be used for different excitation k-space trajectories, for example, k\textsubscript{T} points\textsuperscript{17} for non-selective, spiral/spherical shaped ones for both non-selective and 3D volume selective\textsuperscript{15,16} or spokes for slice selective\textsuperscript{13,14} pTx pulses. For trajectories like spokes or k\textsubscript{T} points, which use only a few pulse shape samples compared to a SPINS trajectory, online RF pulse calculation needs even less time. To further save online computation and measurement time during the sequence preparation phase, pulse shapes could be calculated with lower temporal resolution. Additionally, the applied B\textsuperscript{1+} mapping sequence could be compared to different turbo FLASH protocols\textsuperscript{35} or various B\textsuperscript{1+} mapping sequences (e.g., DREAM or actual flip-angle imaging [AFI]),\textsuperscript{36,37} as also investigated in Chung et al\textsuperscript{35} and Pohmann and Scheffler.\textsuperscript{38} Furthermore, we included extracranial regions into the optimization process, which most likely increased our NRMSE values. Additionally, the fact that IOP\textsubscript{10,1} does not improve NRMSE (Figure S1), and entirely individually optimized FOCUS pulses do not reach lower NRMSE values for their respective subject than other pulses (Figure S4), indicates a general limitation of this approach (limited offline optimization time and number and range of the COV, limitations of the “GlobalSearch” algorithm itself, as well as using a predefined SPINS trajectory and pulse duration). Yet, when using a mask that ignores non-brain tissues in the pTx pulse design, we reached lower mean NRMSE values among all data sets of ~7% (see Supporting Information Figure S6). These values are in good agreement to the NRMSE values.
observed by Gras et al21 for “subject tailored” pulses (individual k-space trajectory and RF shapes). In general, online RF pulse calculation can also be done with several predefined trajectories with corresponding UPs, whereas the most suitable FOCUS pulse for the specific sequence and patient is chosen online. Furthermore, to reduce SAR, the adiabatic inversion pulse could be replaced by a FOCUS pulse as well (using ~78% of this adiabatic pulse’s SAR, data not shown). On the contrary, this would increase online optimization time (from ~15 s to ~70 s online pTx calculation time). In principle, this approach can also be used with a kT point trajectory, although this would limit the available degrees of freedom for online optimization. On the contrary, in this case, the online optimization could use other fast and clinically applicable methods as presented in Guérin et al30 and Hoyos-Idrobo et al39 that more directly use local, global SAR, and power constraints.

In previous works, pTx pulses were applied to smaller study groups of up to 20 subjects to demonstrate the feasibility.8,17,21,24 In this work, we evaluated the design of the pTx pulses on 72 subjects in total and included both patients and different healthy ethnic groups (Europeans and Asians). Compared to CP-mode, the pTx pulses—both UPs and IOPs—showed substantial improvements in FA homogeneity for all subjects despite varying age, head shapes and sizes, as well as anatomical anomalies, which were present in a subgroup of the subjects. Therefore, the designed pTx pulses prove to be robust and thereby clinically applicable. The choice of the regularization parameters such as $w_{\text{Hom}}$ or $w_{\text{SED}}$, or the choice of the pulse type (e.g., IOPs instead of UPs) had a larger influence on NRMSE than the differences between study groups. Pulses derived from different training groups showed similar performances (Figure S3), therefore, dividing the data sets into “healthy Europeans”, “Asians”, and “Europeans with anomalies” might not be required, and other features remain to be explored that allow creating certain clusters having their own optimized COV. As Figure 5 suggests, there is a need for adapting the trajectory and regularization weight to each other, which was done in this work by optimizing the respective parameters simultaneously. For abdominal MRI, Tomi-Tricot et al40 found a larger dependence on the training data. This might be caused by a larger anatomical variability in the abdominal region and also the use of different solvers, pulse calculation techniques and k-space trajectories (SPINS trajectory with individual RF optimization vs. entirely precalibrated UP-based kT-point trajectories).

FOCUS pulses have shown great performance on a large number of subjects. However, there is still potential to further improve it by finding better subject-specific trajectories online in a clinically feasible amount of time. Additionally, the offline optimization process might be improved by using different global solvers, input value ranges, and function and input tolerances, which could potentially lead to COV or UPs that are more tailored to single subjects or groups. Using our solver, we did not observe improved homogeneity or lower SAR values for entirely individually optimized pulses for their respective subjects (see Supporting Information Figure S4). Because the applied solver was used

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**FIGURE 9** MPRAGE images and corresponding FA maps using UP11 and IOP11 of a patient who had brain surgery and is considered to be the subject with the most apparent anomalies in brain tissue. The NRMSE values of UP11/IOP11 were 24.4/14.8 [%], respectively, and the respective SED values were 30.4/50.3 [mJ/kg]. UP11 does still maintain a homogeneous excitation in most parts of the brain. However, there are regions (red arrows) where UP11 shows severe signal loss and leaves parts of anatomic structures invisible. In these regions, IOP11 shows clear improvement, which is visible in both the FA maps and the MPRAGE images.
with a fixed time constraint, it might be possible that our COV and UPs are derived from local minima. Furthermore, using subject-tailored trajectories might be essential for other body parts such as the abdomen where inter-subject variability is higher.  

5 | CONCLUSION

We presented a method to generate FOCUS pTx pulses based on UPs in only ~67 s additional sequence preparation time. Our simulations performed on 72 subjects show, that FOCUS pulses generally reach lower FA-NRMSE values and can also reach a better tradeoff between FA-NRMSE and SAR values than the designed UPs. On 7 additional subjects, we have shown experimentally that, first, UPs and FOCUS pulses improve FA homogeneity compared to CP pulses. Second, the presented FOCUS pulses provide better FA homogeneity than UPs, which is particularly evident in a patient having apparent anomalies.

CONFLICT OF INTEREST

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REFERENCES

1. Pohmann R, Speck O, Scheffler K. Signal-to-noise ratio and MR tissue parameters in human brain imaging at 3, 7, and 9.4 Tesla using current receive coil arrays. *Magn Reson Med*. 2016;75:801-809.
2. Balchandani P, Naidich TP. Ultra-high-field MR neuroimaging. *AJNR Am J Neuroradiol*. 2015;36:1204-1215.
3. Ladd ME, Bachert P, Meyerspeer M, et al. Pros and cons of ultra-high-field MRI/MRS for human application. *Prog Nucl Magn Reson Spectrosc*. 2018;109:1-50.
4. Padorno F, Beqiri A, Hajnal JV, Malik SJ. Parallel transmission for ultra-highfield imaging. *NMR Biomed*. 2016;29:1145-1161.
5. Mao W, Smith MB, Collins CM. Exploring the limits of RF shimming for high-field MRI of the human head. *Magn Reson Med*. 2006;56:918-922.
6. Schmitter S, Wu X, Adriany G, Auerbach EJ, Ugurbil K, Van de Moortele PF. Cerebral TOF angiography at 7T: impact of B1(+) shimming with a 16-channel transceiver array. *Magn Reson Med*. 2014;71:966-977.
7. Schmitter S, Schnell S, Ugurbil K, Markl M, Van de Moortele PF. Towards high-resolution 4D flow MRI in the human aorta using kt-GRAPPA and B1+ shimming at 7T. *J Magn Reson Imaging*. 2016;44:486-499.
8. Metzger GI, Snyder C, Akgun C, et al. Local B1+ shimming for prostate imaging with transceiver arrays at 7T based on subject-dependent transmit phase measurements. *Magn Reson Med*. 2008;59:396-409.
9. van den Bergen B, Van den Berg CA, Bartels LW, Lagendijk JJ. 7 T body MRI: B1 shimming with simultaneous SAR reduction. *Phys Med Biol*. 2007;52:5429-5441.
10. Katscher U, Börnert P, Leussler C, et al. Transmit SENSE. *Magn Reson Med*. 2003;49:144-150.
11. Katscher U, Bornert P, van den Brink JS. Theoretical and numerical aspects of transmit SENSE. *IEEE Trans Med Imaging*. 2004;23:520-525.
12. Zhu Y. Parallel excitation with an array of transmit coils. *Magn Reson Med*. 2004;51(4):775–784.
13. Saekho S, Yip C-Y, Noll DC, et al. Fast-kz three-dimensional tailored radiofrequency pulse for reduced B1 inhomogeneity. *Magn Reson Med*. 2006;55:719-724.
14. Zhang Z, Yip C-Y, Grisson W, et al. Reduction of transmitter B1 inhomogeneity with transmit SENSE slice-select pulses. *Magn Reson Med*. 2007;57:842-847.
15. Hao S, Fessler JA, Noll DC, et al. Joint design of excitation k-space trajectory and RF pulse for small-tip 3D tailored excitation in MRI. *IEEE Trans Med Imaging*. 2016;35:468-479.
16. Davids M, Schad LR, Wald LL, et al. Fast three-dimensional inner volume excitations using parallel transmission and optimized k-space trajectories. *Magn Reson Med*. 2016;76:1170-1182.
17. Cloos MA, Boulant N, Luong M, et al. K-T -points: short three-dimensional tailored RF pulses for flip-angle homogenization over an extended volume. *Magn Reson Med*. 2012;67:72-80.
18. Malik SJ, Keihaninejad S, Hammers A, et al. Tailored excitation in 3D with spiral nonselective (SPINS) RF pulses. *Magn Reson Med*. 2012;67:1303-1315.
19. Cloos MA, Boulant N, Luong M, et al. Parallel-transmission-enabled magnetization-prepared rapid gradient-echo T1-weighted imaging of the human brain at 7 T. *Neuroimage*. 2012;62:2140-2150.
20. Deniz CM, Alon L, Brown R, et al. Subject- and resource-specific monitoring and proactive management of parallel radiofrequency transmission. *Magn Reson Med*. 2016;76:20-31.
21. Gras V, Vignaud A, Amadon A, et al. Universal pulses: a new concept for calibration-free parallel transmission. *Magn Reson Med*. 2017;77:635-643.
22. Gras V, Mauconduit F, Vignaud A, et al. Design of universal parallel-transmit refocusing kT -point pulses and application to 3D T2 -weighted imaging at 7T. *Magn Reson Med*. 2018;80:53-65.
23. Gras V, Pracht ED, Mauconduit F, Le Bihan D, Stöcker T, Boulant N. Robust nonadiabatic T2 preparation using universal parallel-transmit kT -point pulses for 3D FLAIR imaging at 7 T. *Magn Reson Med*. 2019;81:3202-3208.
24. Gras V, Boland M, Vignaud A, et al. Homogeneous non-selective and slice-selective parallel-transmit excitations at 7 Tesla with universal pulses: a validation study on two commercial RF coils. *PLoS One*. 2017;12:e0183562.
25. Kraff O, Fischer A, Nagel AM, et al. MRI at 7 Tesla and above: demonstrated and potential capabilities. *J Magn Reson Imaging*. 2015;41:13-33.
26. Fautz H-P, Vogel M, Gross P, Kerr A, Zhu Y. B1 mapping of coil arrays for parallel transmission. In ISMRM 16. Toronto, 2008.
27. Gerchberg RW, Saxton WO. Practical algorithm for determination of phase from image and diffraction plane pictures. *Optik*. 1972;35:237-246.
SUPPORTING INFORMATION

Additional Supporting Information may be found online in the Supporting Information section.

TABLE S1 Lower bounds (LB) and upper bounds (UB) for COV and optimized values for IOP11 and IOP51 MATLAB’s “Global Search” algorithm generated several potential starting points for local gradient descent algorithms according to initial COV values and the given lower and upper bounds. For practicability we applied a time limit of one week for this optimization.

PSEUDOCODE S1 Pseudocode for the Online SED Regularization Algorithm

FIGURE S1 NRMSE and SED values for different combinations of homogeneity and energy weights ([wHom, wSED] = [1, 2], [1, 1], [2, 1], [3, 1], [5, 1], [10, 1]). Each IOP’s starting COV when solving Eq. 6 are denoted as “xx” start in the index, the starting COV of IOP11 and IOP51 were derived from several successive optimizations. IOP10,1, 51start does not improve NRMSE, although it uses a higher SED than IOP51. For IOP31, 11start or IOP21, 11start there is also no substantial improvement in NMRSE compared to IOP11. IOP21, 51start and IOP31, 51start almost reach NRMSE values as low as IOP51 while using less SED. IOP12, 11start has significantly lower SED values than IOP11 (P = .049) and reaches slightly but also significantly higher NRMSE values (P = 0.015, mean(NRMSE_IOP11): 13.46%, mean(NRMSE_IOP12, 11start): 13.8%, mean(SED_IOP11): 35.9 mJ/kg, mean(SED_IOP12, 11start): 34.4 mJ/kg)

FIGURE S2 Pulse energies of all groups for different pulse types as sum of the squared RF pulse shapes of all channels in volts, divided by their inner resistance of 50 Ω. UP11 uses less pulse energy to excite a higher mean FA than the CP pulse. Scaling up the CP pulse energy to excite the same mean FA as IOP51 (6.9°) would lead to an amount of energy comparable to IOP31 (48.2 mJ)

FIGURE S3 Evaluation of FOCUS pulses with [wHom, wSED] = [5, 1] trained with different sizes of training groups Np on all 72 datasets. Three pulses were trained using a single healthy European subject (“subject1”, “2” and “3”) with two days optimization time, one was trained with 12 subjects and one with 24 healthy European subjects with respective optimization times of one and two weeks. IOP51 is denoted as IOP51, 12 subjects here and outperforms all other IOPs. IOPs trained with single subjects did not show lower cost values for the respective subject they were trained on.

FIGURE S4 Evaluation of FOCUS pulses with [wHom, wSED] = [5, 1] trained with different sizes of training groups Np on all 72 datasets. Three pulses were trained using a single healthy European subject (“subject1”, “2” and “3”) with two days optimization time, one was trained with 12 subjects and one with 24 healthy European subjects with respective optimization times of one and two weeks. IOP51 is denoted as IOP51, 12 subjects here and outperforms all other IOPs. IOPs trained with single subjects did not show lower cost values for the respective subject they were trained on.
outperforms UP\textsuperscript{51} when regularized to its SED bounds, but IOP\textsubscript{51} produces higher NRMSE values than UP\textsubscript{11} when regularized to its SED limits.

**FIGURE S6** Top left: FA maps of IOP\textsubscript{51} of four different subjects both with the used original mask and additional brain extraction. Top right: All slices containing valid pixels of another subject with applied brain extraction mask. Bottom: Various IOPs with \([w\text{Hom}, w\text{SED}] = [5,1]\) trained with different numbers of subjects (same pulses as in Supporting Information Figure S4). All IOPs show mean NRMSE values about 7\% and also lower SED values compared to using no brain extraction (see Supporting Information Figure S4).

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