Three-dimensional static and dynamic parallel transmission of the human heart at 7 T

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Three-dimensional (3D) human heart imaging at ultra-high fields is highly challenging due to respiratory and cardiac motion-induced artifacts as well as spatially heterogeneous $B_1^+$ profiles. In this study, we investigate the feasibility of applying 3D flip angle (FA) homogenization targeting the whole heart via static phase-only and dynamic kT-point in vivo parallel transmission at 7 T. 3D $B_1^+$ maps of the thorax were acquired under free breathing in eight subjects to compute parallel transmission pulses that improve excitation homogeneity in the human heart. To analyze the number of kT-points required, excitation homogeneity and radiofrequency (RF) power were compared using different regions of interest in six subjects with different body mass index (BMI) values of 20-34 kg/m² for a wide range of regularization parameters. One subset of the optimized subject-specific pulses was applied in vivo on a 7 T scanner for six subjects in Cartesian 3D breath-hold scans as well as in two subjects in a radial phase-encoded 3D free-breathing scan. Across all subjects, 3-4 kT-points achieved a good tradeoff between RF power and nominal FA homogeneity. For subjects with a BMI in the normal range, the 4 kT-point pulses reliably improved the coefficient of variation by less than 10% compared with less than 25% achieved by static phase-only parallel transmission. In vivo measurements on a 7 T scanner validated the $B_1^+$ estimations and the pulse design, despite neglecting $\Delta B_0$ in the optimizations and Bloch simulations. This study demonstrates in vivo that kT-point pTx pulses are highly suitable for mitigating nominal FA heterogeneities across the entire 3D heart volume at 7 T. Furthermore, 3-4 kT-points demonstrate a practical tradeoff between nominal FA heterogeneity mitigation and RF power.

KEYWORDS
heart, kT-points, parallel transmission, 7 T

Abbreviations used: 2D/3D, two-dimensional/three-dimensional; BMI, body mass index; CV, coefficient of variation; ECG, electrocardiogram; FA, flip angle; FOV, field of view; GRE, gradient recalled echo; IRB, institutional review board; MR, magnetic resonance; pTx, parallel transmission; RF, radiofrequency; RMS, root mean squared; RMSE, root mean squared error; ROI, region of interest; RPE, radial phase-encoding; SAR, specific absorption rate; SNR, signal-to-noise ratio; SPINS, spiral nonselective; TE/TR, echo time/repetition time; TIAMO, time interleaved acquisition of modes; UHF, ultra-high fields.

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Ultra-high field MR is often limited by inhomogeneities of the radiofrequency (RF) transmit magnetic field ($B_1^+$) yielding spatially variable flip angles (FAs) and, in the worst case, areas with zero FA. Various methods have been proposed to address the problem of spatial heterogeneities of the $B_1^+$ fields, including dedicated coil design,\textsuperscript{2,3} inductively coupled RF resonator arrays,\textsuperscript{3} dielectric padding,\textsuperscript{4} hybrid metasurfaces,\textsuperscript{5} adiabatic RF pulses\textsuperscript{6,7} or, in combination with multiple transmit coils, via parallel transmission (pTx).\textsuperscript{1} Different RF pulse design methods have been developed to compute slice-selective and spatially nonselective pTx RF pulses based on two-dimensional (2D) or three-dimensional (3D) $B_1^+$ maps for multi-dimensional applications, including static $B_1^+$ shimming,\textsuperscript{8} multi-dimensional (pTx-accelerated) excitation,\textsuperscript{9,10} slice-selective spokes,\textsuperscript{11} spatially nonselective kt-points\textsuperscript{12} or spiral nonselective (SPINS)\textsuperscript{13} pulses. Alternatively, precomputed universal pulses can be applied without the need to acquire subject-specific $B_1^+$ maps to improve the $B_1^+$ inhomogeneity in the human heart.\textsuperscript{14}

Cardiovascular MRI (CMR) is generally more challenging because of multiple manifestations of physiological motion, including respiration-induced motion, blood flow and cardiac motion. Despite these difficulties, CMR is used in routine clinical practice at field strengths of 3 or 1.5 T. However, at ultra-high fields (UHF), standard techniques to address the aforementioned challenges, including electrocardiogram (ECG) triggering or the use of respiratory navigators, become less reliable and the RF transmit fields within the target volume becomes increasingly heterogeneous. Nevertheless, the higher SNR available at UHF could be used to push the spatial and temporal limits of cardiac imaging despite the difficulties caused by UHF. One prerequisite to achieving this is a homogeneous $B_1^+$ field. A homogeneous FA in the heart can be obtained using parallel transmission, assuming that the underlying $B_1^+$ field of each transmit coil is known. However, robustly mapping the magnitude and phase of $B_1^+$ in the upper body is a highly demanding task on multi-transmit architectures at UHF. To date, this has primarily been performed with ECG-based cardiac gating in single or multiple breath-holds, limiting $B_1^+$ mapping to a single or a few 2D slices,\textsuperscript{15–17} or alternatively without ECG in free breathing in a single 2D slice.\textsuperscript{18,19}

Various groups have successfully demonstrated pTx based on 2D $B_1^+$ maps in the body, using static $B_1^+$ shimming for smaller 2D regions of interest (ROIs) such as the prostate or the human heart.\textsuperscript{20–22} For larger ROIs, two 2D acquisitions with different $B_1^+$ shims have been applied to address signal nonuniformity by using the time interleaved acquisition of modes (TIAMO) method.\textsuperscript{23} Dynamic pTx with slice-selective spokes RF pulses was shown to further improve excitation uniformity in the liver\textsuperscript{24} and the heart\textsuperscript{15–17} at 7 and 10.5 T.\textsuperscript{25} One main finding of these studies was that static $B_1^+$ shimming is not able to sufficiently mitigate the FA heterogeneity when targeting larger 2D ROIs, and naturally this can become exacerbated when large 3D volumes are targeted.

To date, SAR limits and the various motion sources, including respiration, cardiac motion and blood flow,\textsuperscript{22} have hindered the acquisition of acceptable channel-wise 3D $B_1^+$ maps of the human abdomen at 7 T and have prevented the investigation and development of dynamic 3D pTx pulses to homogenize larger body volumes, as seen at 7 and 9.4 T in the human head\textsuperscript{1,2,26} or at 3 T in the liver.\textsuperscript{27} These limitations can be avoided by the short acquisition of relative or absolute $B_1^+$ maps in one or a few 2D slices performed in a single or multiple breath-holds.\textsuperscript{15,28} For 3D $B_1^+$ maps requiring longer acquisition times, other approaches are needed. Relative 3D $B_1^+$ mapping in the human body while breathing freely\textsuperscript{29} permits a good tradeoff between acquisition time and SAR demands and also allows investigation of the static and dynamic 3D $B_1^+$ shimming in the body.

In this study, we demonstrate the feasibility and benefits of subject-specific spatially nonselective kt-point pulses to achieve 3D FA homogenization across the entire human heart at 7 T. The optimized kt-point pulses were compared with static phase-only $B_1^+$ shimming and were tested via various simulations for multiple kt-points optimized for subject-specific 3D $B_1^+$ maps. In vivo data were acquired with 3D gradient recalled echo (GRE) breath-hold scans in six subjects with different body sizes. Additional data were acquired in two subjects using 3D GRE radial phase-encoding (RPE) in free breathing with subject-specific static $B_1^+$ shimming and dynamic kt-point pulses.

The presented work achieves homogeneous FAs in the human heart using dynamic kt-point pTx pulses based on relative 3D $B_1^+$ maps of the human body and forms the basis for future 3D body imaging applications at UHF.

2 | METHODS

2.1 | Setup and hardware

MRI was performed on a 7 T scanner (Magnetom 7 T, Siemens Healthineers, Erlangen, Germany) equipped with an eight-channel transmit array (1 kW peak power per channel) and a whole-body gradient system, which can achieve a maximum amplitude of 40 mT/m and slew rate of 200 T/m/s. The in vivo measurements were performed with a commercial body coil array (MRI.TOOLS, Berlin, Germany), which consists of 32 transceiver elements (eight dipoles and 24 loops) that are driven in an 8Tx/32Rx channel mode. This coil was certified by a notified body to comply with the local SAR limits in the first level controlled mode of 20 W/kg (IEC 60601-2-33) with a field of view (FOV)/excitation of approximately more than 240 mm along the head-foot direction. Each of the transmit channels consists of one dipole and three loop elements. Reconstruction of the estimated relative $B_1^+$ maps, pulse design and creation of the pulse files was performed on a separate workstation PC (12 cores with 2.1 GHz, 128 GB RAM).
Eight healthy volunteers (four females and four males; average age 29 years, range 21-35 years) with a wide range of body mass index (BMI) values (20-34 kg/m²) were scanned in the supine position with a heart-centered FOV according to an approved institutional review board (IRB) protocol. All subjects participated voluntarily and signed an informed consent form. The volunteers were split into two different groups. Group 1 contained six volunteers whose data were used for a simulation study to analyze different shim settings. Group 2 contained two volunteers who were scanned to qualitatively analyze the subject-specific kT-point pulses in a 3D free-breathing measurement. We did not exclude any volunteer data from this study. The coil consisted of two parts, each with four transmit channels, and were positioned underneath the subject as well as on top of the chest. To monitor vital signs, ECG and infra-red plethysmography were recorded throughout the MR examination. The vital signs were not used in the reconstruction. For each subject, the transmit reference voltage was set to 170 V, which was close to the maximum of the 8 x 1 kW amplifier (Stolberg AG, Stolberg, Germany). The tune-up B₀ shim was used in all but one subject since initial tests showed no clear improvement of the B₀ homogeneity in the heart region after performing the vendor’s B₀ shimming routine. The tune-up B₀ shim was only replaced by a specific B₀ shim for subject 3, who had the highest BMI (34 kg/m²).

For B¹ mapping, a 3D GRE sequence with RPE trajectory was used. Eight 3D GRE datasets were acquired under free-breathing with the full receive capability of the body coil and transmitting via only one transmit channel (ch) while all others were deactivated. In addition, scans with all transmit channels activated, as well as without any transmission, were acquired using the default phase setting, resulting in 10 3D GRE datasets. The default phase was set by the manufacturer to provide sufficient B¹ throughout the aorta. The following parameters were used: nominal FA = 20°; TE/TR = 2.02/40 ms, FOV = 250 × 312 × 312 mm³, slice thickness = 4 mm, bandwidth = 399 Hz/Px and 256 RPE lines separated by the golden angle, resulting in a total acquisition time of 205 seconds to acquire all 10 3D GRE datasets. For subject 3, an extended FOV = 250 × 350 × 350 mm³ with otherwise identical parameters was used. Following acquisition, the raw data were exported to the remote workstation and reconstructed for an isotropic voxel size of 4 mm in less than 1 minute using full k-space information and without correction of respiratory or cardiac motion.

Relative 3D channel-wise B¹⁺ estimates were computed using the 3D GRE images from each transmit channel assuming that the sum of magnitudes of all transmit channels is equal to the sum of magnitudes of all receive channels for each spatial position of the eight channels. Figure S1 contains the source code to compute the channel-wise relative B¹⁺ maps. The underlying method has been frequently applied in 2D acquisitions providing good results for different 2D applications in the human body at 7 T. The channel-wise relative phase distributions were calculated relative to the first transmit channel. The channel-wise superpositions of the resulting B¹⁺ maps were used to manually draw the ROI of the heart on a slice-by-slice basis for each subject. This took less than 1 minute for 9-16 slices. The ROI was used as a binary mask for pulse design and quantitative evaluation.

ΔB₀ maps were acquired in 21 seconds using a bandwidth of 1260 Hz/Px, three different echo times (Tₑ1/Tₑ2/Tₑ3 = 1.02/2.04/3.06 ms, respectively) and a bipolar readout on an RPE trajectory. All other parameters were identical to those used for B¹⁺ mapping. Using the bipolar readout, the amplitude and phase of data of Tₑ2 were corrected prior to reconstruction. Finally, ΔB₀ maps were calculated with a multi-seeded region-growing algorithm.

2.3 Pulse design

In this work, we investigated the feasibility of improving the excitation homogeneity throughout the human heart using static phase-only RF shimming and dynamic magnitude and phase pTx for multiple kT-points with the small-tip-angle approximation. The assumption of small FA allowed manual calibration of the relative B¹⁺ maps to compute the nominal FA assuming a linear relationship between B¹⁺ and the FA. The scaled B¹⁺ maps were then used for the pulse design to predict and optimize the nominal FA distribution in each ROI. Note that this does not reflect the actual FA.

Three sets of ROIs were used to optimize the pulses:

I. ROI₁: a single slice of ROI₄ in the isocenter.
II. ROI₂: three slices of ROI₄ symmetrically distributed around the isocenter with a slice gap of 8 mm.
III. ROI₃: heart ROI covering 9-16 slices with a slice gap of 8 mm (on average 12 slices).

All pulse designs were computed in MATLAB R2014b (MathWorks, Natick, MA, USA).

2.3.1 Static phase-only shim

Static phase-only shimming was performed similarly to by solving a cost function that optimized the tradeoff between homogeneity measured by the coefficient of variation (CV)
Optimization was based on the magnitude of the $\hat{B}_1^+$ maps superimposed with complex RF phase factors $b_{ch} = e^{j\phi_{ch}}$, with $\phi_{ch}$ being the channel-dependent phase-offset and $N_c$ being the number of channels and the sum of the magnitude of the $\hat{B}_1^+$ maps for a given ROI. Each optimization was performed eight times (this number was set empirically to avoid local minima) with different pseudo-random starting phases followed by an automated selection of the best solution based on the lowest cost function value. The computation times were of the order of seconds.

### 2.3.2 Dynamic kT-points

Dynamic kT-points consist of a series of complex weighted rectangular RF pulses separated by gradient blips to homogenize 3D volumes. The pulse design problem was solved using the small-tip-angle approximation with an interleaved greedy and local optimization solving

$$
\min_{b_{ch}} \frac{1}{2} \| m - \sum_{ch=1}^{N_c} \hat{B}_{1,\text{ch}} A(K) b_{ch} \|_{\text{ROI}}^2 + \frac{\beta}{2} \| b \|^2,
$$

(3)

to compute the complex RF weights $b_{ch}$ (magnitude and phase) and k-space locations $K$ for each kT-point, with $m$ being the desired target pattern or FA, $\hat{B}_{1,\text{ch}}$ being the estimated $\hat{B}_1^+$ maps for each channel $ch$, $A(K)$ being the excitation system matrix and $\beta$ being the regularization term balancing excitation fidelity and RF power, as described in Grissom et al. The complex RF weights $b_{ch}$ were computed by solving the magnitude least-squares solution followed by a greedy method to select the next kT-point using. This alternating update scheme was repeated for all kT-points without coil compression for all channels. The initial regularization parameter was updated every 50 iterations, as described in Grissom et al. The temporal resolution of the RF pulses and gradient blips was set to the gradient raster time of 10 ms. Again, voxels outside the ROI were ignored by a binary mask. The impact of $\Delta B_0$ variations on the optimized kT-point pulses was analyzed for subject 1 using $\Delta B_0$ maps obtained with an RPE-based 3D multi-echo GRE sequence. If not otherwise stated, all other kT-point optimizations have been computed without including a $\Delta B_0$ map.

### 2.4 Qualitative analysis

Two different excitation settings (a) static phase-only shim optimized for ROI3 (ROI3 static shim) and (b) 4 kT-points optimized for ROI4 (ROI4, 4 kT-points) were acquired in six healthy volunteers (group 1) to qualitatively check the computed $\hat{B}_1^+$ estimates $\hat{B}_{1,\text{ch}}$. The rectangular (hard) RF pulse series with optimized phase shim (ROI4 static shim) and the kT-points RF pulses and interleaved gradient blips were inserted in a high-resolution Cartesian 3D GRE sequence. The subjects were scanned in the supine position. To achieve abdominal coverage in the anterior-posterior (A-P) direction for all subjects, the amount of phase-encoding lines was adapted from 56.3% up to 75% (216 up to 288 mm). The increase in phase-encoding lines was compensated for by a reduction of the phase resolution to 70% to achieve a similar scan duration of ~30 seconds for each subject. Aside from these subject-specific adaptations, all the other parameters remained constant: nominal FA = 30°, TE/TR = 1.3/3.4 ms, FOV read = 384 mm, base resolution = 320, slice thickness = 2.77 mm, slices = 72, slice oversampling = 77.8% and GRAPPA = 2 with 24 reference lines and a bandwidth of 1120 Hz/Px. The short scan duration allowed acquisition of the entire 3D volume in a single breath-hold in the end expiration state.

### 2.5 Quantitative analysis

The impact of multiple kT-points, regularization parameters and ROIs on excitation fidelity, as well as the required RF power and pulse duration, were analyzed in a simulation study following acquisition of six in vivo subjects from group 1.

Two different timing schemes, (a) fixed subpulse duration and (b) fixed total pulse duration for adding kT-points, were implemented, and kT-points were optimized using fixed and automatic (updated every 50 iterations) regularization parameters. The first optimization runs were

\[ CV = \text{std} \left( \left| \sum_{n=1}^{N_c} B_{1,\text{ch}} b_{ch} \right|_{\text{ROI}} \right) / \text{mean} \left( \left| \sum_{n=1}^{N_c} B_{1,\text{ch}} b_{ch} \right|_{\text{ROI}} \right) \]
computed for a fixed subpulse duration (a) of 0.24 ms, which consisted of a 0.1 ms square RF pulse followed by a 0.14 ms gradient blip. Each additional kT-point increased the total pulse duration by 0.24 ms, resulting in total pulse durations of between 0.48 and 1.44 ms. The second optimization runs were computed to achieve a fixed total pulse duration (b) of 0.96 ms with variable square RF pulse durations of between 0.36 and 0.02 ms. Each optimization was performed 20 times with different pseudo-random starting RF phases.

Static phase-shims and dynamic pTx pulses with 4 kT-points were both optimized for ROI1, ROI3 and ROIH. The optimized RF weights and kT-points were then converted to RF and gradient vectors. Dynamic kT-points were implemented using four 0.1 ms-long square RF pulses separated by 0.14 ms-long gradient blips, leading to a total duration of 0.96 ms. To achieve comparability, default phase settings and static phase-only shim settings were implemented with the same structure, except that the blip moments were set to 0 and the same phases were applied to all 4 kT-point pulses. Default and static shim RF vectors were also scaled to achieve the same nominal mean FA over the ROI as for the optimized kT-points pulse. The CV was computed within different ROIs to confirm the validity of the RF and gradient vectors prior to insertion into the MR sequence.

2.6 Experimental validation

Two subjects (group 2) were additionally scanned during the same MRI session after relative 3D $B_1^+$ mapping and kT-points pulse design with a high-resolution 3D RPE-GRE free-breathing sequence using 4 kT-points excitation pulses optimized for ROIH. The following parameters were used: nominal FA = $10^\circ$, TE/TR = 1.75/3.7 ms, FOV = $250 \times 312 \times 312$ mm$^3$, slice thickness = 1.4 mm, bandwidth = 1015 Hz/Px and 256 RPE lines$^{30}$ separated by the golden angle, resulting in a total acquisition time of 333 seconds. The acquired 3D RPE dataset was binned into four

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**FIGURE 1** Relative 3D estimated $B_1^+$ maps (left: magnitude, right: phase angle) of subject 1 at 7 T. Four channels are located on the chest and four channels are positioned under the subject’s back. The phase angle of channels 2 to 8 are computed relative to channel 1.
respiratory motion states using self-navigation and was reconstructed for an isotropic voxel size of 1.4 mm with a respiratory motion surrogate retrieved from a one-dimensional (1D) projection in the head–feet direction in the k-space center. The underlying motion field was estimated via image registration based on the aforementioned reconstructions and used to correct the respiratory motion. The resulting respiratory-corrected 3D dataset was then used to qualitatively validate the $B_1^+$ predictions.

3 | RESULTS

3.1 | $B_1^+$ maps and manually drawn ROIs

Figure 1 shows the relative 3D $B_1^+$ maps (4-mm isotropic magnitude and phase) of subject 1 for eight transmit channels in transversal, coronal and sagittal views. As clearly indicated by the maps, channels 4–7 are located posteriorly, while the remaining channels are located anteriorly. To suppress noise, the $B_1^+$ maps were masked based on a threshold applied to the sum-of-squares image of the 32 receive channels. Both the magnitude and the phase of each channel are free of breathing artifacts, despite acquiring the underlying 3D GRE data under free breathing. The same observation was made for all eight subjects. The 3D $B_1^+$ maps are free of motion artifacts and show magnitude and phase distributions comparable with the 2D reference images, thereby justifying the use of nonrespiratory-resolved 3D $B_1^+$ maps.

Figure 2 shows multiple transversal slices of the magnitude of the complex sum of $B_1^+$ maps shown in Figure 1, reflecting the default phase excitation mode. The resulting maps yield a strong $B_1^+$ gradient along the A-P direction across the heart and a band of low $B_1^+$ along the left–right (L-R) direction across the heart. Twelve 2D ROIs were drawn manually on a slice-by-slice basis to define the subject-specific volume ROI$_H$. The voxels outside the ROIs were subsequently excluded by applying a binary mask.

3.2 | Pulse design

Figure 3 shows the RF and gradient vectors that were selected and applied for the experimental validation in subject 1. These include (a) a default phase setting, (b) the optimized static ROI$_S$ phase-only shim, and (c) the optimized dynamic ROI$_H$ 4 kT-point pTx pulses.

Figure 4A shows the resulting nominal FA maps of subject 1 obtained from the Bloch simulations for the pulses shown in Figure 3. Note that the nominal FA results from a manual scaling of the RF pulses with respect to the relative $B_1^+$ maps and therefore does not reflect an absolute $B_1^+$.
measure. Figure S2 additionally shows the binary mask ROI₄ and results for all three of the different target regions, ROI₁, ROI₃ and the entire heart volume (ROI₄), for static shim and 4 kT-points. For visualization purposes, a shaded mask has been added that outlines the body shape (not part of the optimization). The static shim reduced the CV to 14.5% for ROI₁ and ROI₃ and to 21.6% for ROI₄. As expected, the 4 kT-point pulses performed best with a CV of ~5.6% for ROI₁ and ROI₃ and ~8.5% for ROI₄. Similar values have been observed for all but one of the eight subjects. Figure 4B shows the simulated FAs of subject 1 in the L-R and A-P directions with default, static shim ROI₃ and 4 kT-point ROI₄ pulses for three representative slices. Figure 4C shows the corresponding histograms for ROI₄. In all slices the 4 kT-points shim yields a homogeneous nominal FA, in contrast to the default and static shims, which show substantial spatial nominal FA variations. The standard deviation of the nominal FA for the default shim (9.6 ± 4.3%) is dramatically reduced across ROI₄ by the optimized static phase-only shim (9.5 ± 2.2%) and the proposed 4 kT-point pTx pulses (10.0 ± 0.9%).

Figure 5 shows measured ΔB₀ maps and simulated nominal FA maps with and without the inclusion of the ΔB₀ maps in the kT-point pulse designs and Bloch simulations. The ΔB₀ maps, acquired with the tune-up shim, show ΔB₀ variations ranging between −335 and −4 Hz (95 percentile) with a median of −178 Hz across the ROI₄. The impact of excluding ΔB₀ in the pulse design for the experimental validation was analyzed by the nominal FA results of three simulations: (I) ΔB₀ is excluded in both the kT-points pulse design and Bloch simulation (first line), (II) ΔB₀ is excluded in the kT-points pulse design but included in the Bloch simulation (second line), and (III) ΔB₀ is included in both the kT-points pulse design and Bloch simulation (third line). The nominal FA homogeneity remains almost unchanged by including ΔB₀ in the simulations with a CV of 8.5% (case I) and 8.7% (case II). Inclusion of ΔB₀ in the optimization and simulation resulted in a CV of 8.4% (case III).

3.3 | Quantitative analysis

Figure 6 shows double-logarithmic L-curve plots of the root mean squared (RMS) RF power and the nominal FA root mean squared error (RMSE) for ROI₄ of subject 1 (Figure 6A,B) and the final results for all six subjects in group 1 (Figure 6C). Two sets of optimizations with a fixed subpulse duration (Figure 6A) and fixed total pulse duration (Figure 6B) show the impact of adding kT-points upon RF power and nominal FA homogeneity. Figure 6A shows increasing pulse performance when adding more kT-points. However, this reduction comes at the cost of a linearly increasing total pulse duration since each additional kT-point adds 0.24 ms to the total pulse duration. Figure 6B shows the optimized results with a fixed total pulse duration of 0.96 ms. In this case, increasing the number of kT-points comes at the cost of shorter RF subpulses, which strongly impacts RF power. The results obtained with the automatic regularization parameter are highlighted in Figure 6A,B. Figure 6C shows the results for all subjects using an automatic regularization parameter and fixed total pulse duration of 0.96 ms. The highlighted data points represent the 4 kT-point settings used in the experimental validation. For all six subjects, the choice of 3-4 kT-points resulted in a good tradeoff between excitation fidelity and RF power requirements. The optimized gradient blips required maximum slew rates of 50-100 T/m/s across all subjects. The
FIGURE 4

A. 3D view of the nominal FA maps after Bloch simulation of the complex sum of estimated $B_1^+$ maps with the default phase setting (default), optimized ROI3 static phase setting (static shim) and optimized dynamic entire heart volume (ROIH) 4 kT-points of subject 1. The nominal FA is a result of the manually scaled relative $B_1^+$ maps assuming a linear relationship between $B_1^+$ and the FA. Note that this assumption only holds for small FAs. The Bloch simulations of the optimized static and dynamic pulses for ROI1, ROI3 and ROIH are shown in Figure S2. B, comparison of the simulated nominal FA patterns across the heart for the three shim settings used in the experimental validation of subject 1: default, static shim ROI3 and 4 kT-points ROIH. Depicted are the two outermost slices and the central isocenter slice of the manually drawn ROIH indicating the position of the 1D lines. For each slice, two 1D lines in the right–left and anterior–posterior directions are depicted. The homogeneous nominal FA achieved by the proposed 4 kT-points also holds for the slices not shown. C, the nominal FA distribution is depicted in histograms for the default, static shim ROI3 and 4 kT-points ROIH, demonstrating the superior nominal FA homogenization with the 4 kT-points for ROIH.
computation time to design 6 kT-points for the largest ROI was less than 30 seconds and less than 10 seconds for 4 kT-points using the automatic regularization parameter.

Figure 7 shows a side-by-side comparison of the CV for the default, static phase-only and dynamic 4 kT-point shims respectively optimized for ROI1, ROI3 and ROIH. With the exception of subject 3 (highest BMI), where the 4 kT-points shim of ROIH yielded a CV of 16.5%, the use of 4 kT-points resulted in a CV of less than 10% in each of the remaining five subjects in group 1. By comparison, static phase-only shims yielded an average CV of 25% across all subjects.

Figure 8 shows a graphic visualization of all six 3D $B^*_1$ predictions for group 1 using the calibrated relative $B^*_1$ maps with the ROIH optimized subject-specific 4 kT-points analyzed quantitatively in Figure 7. With the exception of subject 3, where a slight drop in the nominal FA at the top and the back part of the heart is visible, the $B^*_1$ predictions for the remaining subjects show a smooth nominal FA distribution across the whole heart. The predictions also show that nearby tissue such as the aortic arch or the descending aorta also benefit from the kT-points shim with smooth nominal FA patterns, although these regions were not part of the optimization.

### 3.4 Qualitative analysis

Figure 9 shows a qualitative analysis of $B^*_1$ predictions based on the manually calibrated relative $B^*_1$ maps with the default phase setting, optimized static phase-only and dynamic 4 kT-point pulses for subject 1. Residual signal drops apparent in the coronal and sagittal views using the static phase-only shim, indicated by the arrows, were fully compensated for by the 4 kT-points. Overall, compared with the default and optimized phase setting, the 4 kT-points resulted in a clear improvement of the nominal FA homogeneity throughout the ROIH, despite the larger ROI used for kT-optimization. The side-by-side comparison of the simulated and measured default and static shim settings for subjects 2–6 are shown in Figures S2, S3 and S4. Again, the experimental GRE measurements reproduce the FA patterns predicted by the Bloch simulations.
Figure 10 shows the acquired, respiratory-corrected 3D GRE images of subjects 7 and 8 (group 2) using subject-specific optimized 4 kT-point pulses for ROIH. Note that the shown 3D data have been acquired without cardiac gating due to scan time limitations resulting in unresolved cardiac motion. Nevertheless, there is an excellent match between $B_1^+$ predictions and the 3D GRE images, demonstrating the feasibility of achieving 3D spatial excitation uniformity with 4 kT-points. The remaining signal drop in the A-P direction results from the $B_1^-$ receive profile, which has not been corrected in the plots.
FIGURE 8  Simulated nominal FA predictions for subjects 1 to 6 based on the manually scaled relative B1\(^+\) estimations and the dynamic 4 kT-point pulses optimized for the entire heart volume (ROI\(_h\)). Depicted are three views of the 3D volume to demonstrate the nominal FA homogenization across the entire human heart using the proposed 4 kT-point pulses.

FIGURE 9  Side-by-side comparison of B1\(^+\) predictions and reconstructed 3D breath-hold images (exhale) with the default phase setting as well as the optimized static phase-only and dynamic 4 kT-point pTx pulses of subject 1. Depicted are three views of the acquired 3D volume close to the position of the simulated views. The arrows point to signal drop-out regions in the heart, depending on the RF shim. The optimized 4 kT-point pulse corrects for B1\(^+\) variations and consistent signal is achieved across the entire heart volume (ROI\(_h\)). The remaining signal changes in the A-P direction of the acquired data are a result of receive (B1\(^-\)) variations. The comparison of the default phase setting, optimized static phase-only setting and 4 kT-points for the remaining subjects 2–6 are shown in Figures S3, S4 and S5.
In this study, mapping of the relative 3D $B_1^+$ fields was performed using an RPE-based GRE sequence during free breathing in 3 minutes 25 seconds for all eight transmit channels (4-mm isotropic resolution) utilizing a small FA approach. This approach was chosen over other relative $B_1^+$ mapping approaches because of two main advantages. First, it has a short acquisition time and, second, the $B_1^+$ estimation is robust against motion or flow due to short echo times. However, by providing only relative and not absolute $B_1^+$ maps, the resulting maps are biased by the square root of the proton density and by the assumption that the sum of magnitudes of all transmit channels is approximately equal to the sum of magnitudes of all receive channels. Therefore, we expect deviations to the actual $B_1^+$ field, particularly within close proximity to the coil.

Nevertheless, the same approach, which was used here to obtain 3D $B_1^+$ maps, has been successfully applied in several different 2D applications in the human body at 7 T, including generation of $B_1^+$ shimming solutions as well as spokes pTx pulses for the heart. The employed 3D $B_1^+$ mapping method was tested with a motion phantom and was compared with 2D (slice-selective) $B_1^+$ mapping (not shown). We further validated the nonrespiration-resolved and respiration-resolved 3D $B_1^+$ mapping method acquired in free breathing with reference 2D $B_1^+$ maps acquired in a breath-hold. To simplify the practical handling of the experiments, and because of only minor differences between nonrespiration-resolved and respiration-resolved $B_1^+$ maps in the present case of shallow breathing, respiration was not resolved in this work. Nevertheless, the resulting $B_1^+$ maps were free of obvious motion artifacts in all subjects and the high-resolution GRE scans with the optimized pulses fit well to the $B_1^+$ predictions. Preliminary data suggest that using the nonrespiration-resolved 3D $B_1^+$ maps to perform static or dynamic pTx is only justified for shallow breathing, which was the case in all subjects in the current work. For breathing patterns with a larger respiratory amplitude, the RF pulses calculated on the nonrespiration-resolved $B_1^+$ maps appear to perform inferiorly over the entire breathing cycle compared with respiration-robust pulses calculated based on multiple respiration-resolved $B_1^+$ maps.

Relative $B_1^+$ mapping, manual ROI selection and pulse design were performed in less than 10 minutes while the subjects remained in the MR scanner. To minimize this time, dynamic kT-point pulse design was carried out with an automatic regularization term with the small-tip-angle.
approach utilizing the spatial domain method, a justifiable simplification for low-FA rapid GRE imaging with a desired FA of 10°. If larger tip angles were required, the Bloch equations would have to be solved numerically for each subject, resulting in much higher computational effort. It has been shown that similarities in the $B_1^+$ maps of different subjects can be exploited to compute universal pulses, thus avoiding the necessity for subject-specific pulse design and long computation times during the scan. While this was shown for brain imaging, the $B_1^+$ maps of the human abdomen demonstrate a much higher inter-subject variability, thus making it more difficult to apply the universal pulses concept. This is supported by observed variations of the 3D $B_1^+$ maps for different BMI values and body shapes, resulting in different constructive/destructive interference of the different channels. An alternative approach may be to utilize machine learning for fast RF pulse design as proposed for single and parallel applications to the human head. Large FA pulses can be valuable for cardiac imaging to introduce tissue contrast, although high $B_1^+$ peak and RF power are limited in the body at UHF. Note that for large FA pulse design, absolute $B_1^+$ information is required for accurate FA prediction. The manual scaling of the relative $B_1^+$ maps used in this study only works for the small-tip-angle regime, where a linear relationship between $B_1^+$ and nominal FA can be assumed.

Another extension of the presented kT-points pulse design could be to include local SAR instead of RF power constraints. Here, we used power constraints because the RF power per channel is limited to fulfill the safety guidelines. Local SAR supervision was based on a worst-case shim analysis using electromagnetic field simulations of the DUKE model with 10 million random phase settings. The worst-case shim resulted in a $10 \text{ g-averaged peak spatial SAR of } 11.1 \text{ W/kg}$, thus leaving an additional safety margin of 1.8 to the 20 W/kg limit. This approach, that is, choosing the worst shim plus limiting the RF power per channel and setting an additional safety factor, is expected to be more restrictive than monitoring with virtual observation points (VOPs), which are frequently used for pTx applications in the human brain. Thus, we decided in favor of this approach, as our own calculation demonstrated that it yields peak SAR values that are less sensitive to the body shape and size compared with supervision using VOPs.

The acquisition of the absolute $B_1^+$ field or the actual FA for a 3D volume in the human body at 7 T is a very demanding task. Besides the limitations set by peak and mean RF power, scan times, respiratory and cardiac motion, as well as blood flow, the $B_1^+$ mapping method requires a high dynamic range, as the $B_1^+$ fields rapidly drop when moving from the surface towards the center of the body. Despite experimental difficulties, actual FAs in the experimental validation have been estimated retrospectively with a 3D actual flip-angle (AFI) approach. Multiple groups have proposed designing kT-points in the large-tip-angle regime, including applications to the human head. Large FA pulses can be valuable for cardiac imaging to introduce tissue contrast, although $B_1^+$ peak and RF power are limited in the body at UHF. Note that for large FA pulse design, absolute $B_1^+$ information is required for accurate FA prediction. The manual scaling of the relative $B_1^+$ maps used in this study only works for the small-tip-angle regime, where a linear relationship between $B_1^+$ and nominal FA can be assumed.

In contrast to a 2D slice-by-slice shim, where individual phase settings are computed for each slice, we optimized all static and dynamic pulses for one 2D and two 3D ROIs. Compared with a static phase-only $B_1^+$ shim, the higher degrees of freedom of 4 kT-point pulses yielded more homogeneous excitation patterns at the cost of higher RF amplitude and RF power requirements across all subjects and ROIs. For smaller volumes (ROI1 and ROI3), the CV of optimized static and dynamic shims showed a similar reduction across all subjects. In the case of ROI16, the optimized 4 kT-point pulses resulted in a CV of less than 10% in all but one subject. For this subject, the 4 kT-point pulses brought a smaller reduction of homogenization performance, possibly as a result of the higher BMI and thus higher RF power demands compared with the other subjects. However, even in this case the homogeneity of the dynamic $B_1^+$ shim outperformed the static $B_1^+$ shim and the overall image quality agreed well with the Bloch simulations. It should be noted that the nonconvex nature of static and dynamic $B_1^+$ shim optimization does not guarantee finding the global optimum.

A relatively short pulse duration of 0.96 ms was used to limit the sensitivity of the optimized kT-point pulses to $B_0$ off-resonance effects. This approach was supported by Bloch simulations showing no significant changes in the FA maps (CV = 8.5% vs. 8.7%) when pulses designed for $\Delta B_0 = 0$ were simulated with and without the presence of true $B_0$ inhomogeneities. Therefore, it was not deemed necessary to measure the $\Delta B_0$ maps for each subject and to include them in the pulse design. The optimized pTx pulses were integrated into a 3D Cartesian GRE sequence where data were acquired in a single breath-hold, and a 3D RPE GRE sequence where data were acquired during free breathing. Both scans were reconstructed with unresolved cardiac motion. Although the data agree well with the unresolved cardiac motion of the $B_1^+$ maps, clinically relevant cardiac images require resolved cardiac motion. Further important steps towards clinically relevant cardiac MRI are the generation of clinically valuable contrast, a limitation of the presented small-tip-angle method and the evaluation of potential SNR gains when using kT-points compared with static pTx.

The experimental scans were performed to validate $B_1^+$ mapping and different pulse design methods and not to generate high-quality heart images at 7 T. For this purpose, the integration of kT-point pulses into cardiac-resolved 3D sequences will be analyzed in the future.

This study aimed to demonstrate the experimental feasibility and benefits of applying 3D kT-points excitation to the human heart at 7 T. The presented 4 kT-points approach resulted in a FA CV of ~10% across all six subjects compared with ~25% achieved with static phase-only pTx. Therefore, subject-specific kT-points could form the basis for future 3D imaging applications in the body, such as the heart or liver, and have the potential to push the limits of body imaging at 7 T and above.
Mitigating nominal FA heterogeneities across the entire 3D heart volume at 7 T was made feasible by kT-point pTx pulses designed with subject-specific 3D $B_1^+$ estimations. Across a wide range of subjects with different body shapes, 3-4 kT-points demonstrate a practical tradeoff between nominal FA heterogeneity mitigation and RF power.

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