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Cardiac Imaging at 7T: Single- and Two-Spoke RF Pulse Design with 16-channel Parallel Excitation

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Abstract

Purpose—Higher SNR and improved contrast have been demonstrated at Ultra-high magnetic fields (7T) in multiple targets, often with multi-channel transmit B_1^+ methods to address the deleterious impact on tissue contrast due to spatial variations in B_1^+ profiles.

When imaging the heart at 7T, however, respiratory and cardiac motion, as well as B_0 inhomogeneity, greatly increase the methodological challenge. In this study we compare 2-spoke parallel transmit (pTX) RF pulses with static B_1^+ shimming in cardiac imaging at 7T.

Methods—Using a 16-channel pTX system, slice-selective 2-spoke pTX pulses and static B_1^+ shimming were applied in cardiac CINE imaging. B_1^+ and B_0 mapping required modified cardiac triggered sequences. Excitation homogeneity and RF energy were compared in different imaging orientations.

Results—2-spoke pulses provide higher excitation homogeneity than B_1^+ shimming, especially in the more challenging posterior region of the heart. The peak value of channel-wise RF energy was reduced, allowing for higher flip angle, hence increased tissue contrast. Image quality with 2-spoke excitation proved to be stable throughout the entire cardiac cycle.

Conclusion—2-spoke pTX excitation has been successfully demonstrated in the human heart at 7T, with improved image quality and reduced RF pulse energy when compared to B_1^+ shimming.

Introduction

Recent advances in ultra-high field (UHF) MRI operating at field strengths of 7T have demonstrated significant improvements in contrast and image quality for a variety of applications at 7T (1), in particular for vascular targets (2, 3) that directly benefit from higher SNR and longer T_1 relaxation times at UHF (4, 5).

Cutting edge UHF technology has already enabled initial studies of cardiac and coronary imaging at 7T (4, 6–13). The heart, however, is a challenging organ for MRI due to cardiac and respiratory motion, requiring cardiac gating and/or acquisitions performed under breath hold. Additionally, at UHF the shorter RF wavelength compared to lower field strength

leads to strong spatial heterogeneities of the transmit B_1^+ magnitude (14, 15), resulting in contrast variations and potentially severe signal losses in the human heart at 7T (6, 9).

To address this problem, multi-channel transmit (TX) coils in combination with B_1^+ shimming (16–18) have been successfully applied in multiple targets (19, 20), including the heart (4, 6–8, 21). In this study we follow the often used definition of B_1^+ shimming (also labeled “static B_1^+ shimming”) as an approach where each of the individual transmit channels of a coil array is fed by a common RF pulse shape with a complex, channel specific factor controlling RF phase and/or amplitude, resulting in spatially more uniform flip angles. In general, excitation non-uniformity has also been addressed with parallel transmission (pTX) (22, 23) which in contrast to B_1^+ shimming allows the complex, channel-specific factors to vary temporally during the course of an RF pulse, providing additional degrees of freedom to improve excitation homogeneity and/or reduce total RF energy, as shown in the brain or liver (24–27). And a simulation study (28) has indeed shown that cardiac imaging is also expected to directly benefit from multi-spoke pTX RF pulses at 7T.

However, performing cardiac pTX acquisitions at 7T is experimentally fairly challenging due to heart motion, respiratory motion and blood flow. Heart motion requires cardiac triggered acquisition for both B_0/B_1^+ calibration maps as well as for the pTX sequence. However, ECG recording at 7T is significantly impacted by the magneto-hydrodynamic effect as demonstrated by Suttie et al. (7), and a careful placement and potential relocation of the ECG leads is required in order to obtain a suitable ECG signal. Using pTX resources also requires precise synchronization between the individual TX channels and sequence execution for cardiac-gated MR sequences. Furthermore, for the pTX application the question arises as to whether a single B_1^+ solution can be suitable for CINE imaging over the entire cardiac cycle. To avoid respiratory motion artifacts in calibration data and pTX acquisition it is desirable to perform all acquisitions under breath-hold, which means that a robust multi-channel B_1^+ calibration should ideally be completed within a mere period of about 20s, a challenge even more demanding as the number of transmit channels increases. Limiting the impact of blood flow effects also requires designing pTX RF pulses as short as possible.

In this work we investigate the feasibility of applying slice selective 2-spoke pTX RF pulses in cardiac CINE imaging at 7T using a 16-channel pTX system, in the context of the aforementioned constraints. We acquire B_1^+ calibration maps, using a very fast, multi-channel B_1^+ calibration approach, as well as off-resonance frequency maps (ΔB_0 maps), required for the RF pulse design, both collected under breath-hold and with ECG triggering. We compare the resulting flip angle homogeneity and RF energy of the 2-spoke excitation with B_1^+ shimming results.

Methods

Principle and Optimization

In the small tip angle approximation, the transverse magnetization $m(\mathbf{r})$ generated by an RF excitation can be written by the Fourier integral of the excitation k-space trajectory weighted by the sum of the spatial sensitivities $B_{1,i}^+(\mathbf{r})$ of the K individual TX channels and the RF pulses $b_i(t)$ applied to each channel:

$$m(\mathbf{r}) = i\gamma m_0 \sum_{i=1}^K B_{1,i}^+(\mathbf{r}) \int_0^T b_i(t) e^{i\gamma \Delta B_0(\mathbf{r})(t-T)} e^{i\mathbf{k}(t)\mathbf{r}} dt \quad [1]$$

In the discrete case this equation can be represented by a matrix formulation, which is the basis for the spatial domain method for RF pulse design as demonstrated by Grissom et al (29):

$$m = Ab \quad [2]$$

In this formulation, the vector b with length $K \times L$ contains the RF waveforms of the K TX channels for L discrete time points, and the system matrix A incorporates the excitation trajectory, spatial maps of magnetic field deviations ΔB_0 and the transmit B_1^+ sensitivity maps.

In this work we aim to obtain slice selective cardiac CINE images with improved in-plane excitation homogeneity. This can be achieved using 3D-tailored RF pulses (24, 30, 31), so called ‘spoke pulses’, which deposit RF energy in excitation k -space along line segments (spokes) in k_z direction while the individual spokes are separated by shifts in k_x and k_y direction. Due to its intrinsic slice selectivity, the technique requires only 2D sensitivity maps and 2D ΔB_0 maps for the slice of interest (instead of 3D maps), which is advantageous for B_0/B_1^+ mapping requirements and also reduces significantly the optimization complexity.

The goal of the optimization process is to find a suitable vector b that produces a transverse magnetization m which matches closely a desired target magnetization m_t . To solve this problem, we utilize in this work the magnitude least squares optimization (32):

$$\hat{b} = \argmin(\| |Ab| - |m_t| \|_w^2 + \lambda \|b\|^2) \quad [3]$$

The algorithm minimizes the difference between the magnitudes of desired and actual excitation pattern within a predefined ROI (see e.g. Figure 1+2) while limiting the total RF energy by the regularization term $\|b\|^2$, and the tradeoff is controlled by the regularization parameter λ . Pixels located outside the ROI are ignored by the algorithm through the binary matrix w . In this work the entries of vector m_t are set to the real value 1 and according to the linearity of the small tip angle approximation the RF voltages of the K TX channels are scaled to achieve flip angles of about 10° in the experiment. Within the optimization the individual spokes are treated as hard pulses, thus the vector b in equation 3 has the length $K \times S$ and contains the complex weighting for the K channels and S spokes. In practice, equation 3 is solved iteratively using the least squares optimization $\| |Ab| - |m_t| \|_w^2 + \lambda \|b\|^2$, while updating the phases of the target magnetization m_t as described in (32).

In this work we compare the performance of B_1^+ shimming and 2-spoke excitation for cardiac imaging at 7T. Due to technical limitations no more than 2-spokes will be used for the pulse design. A single spoke excitation ($S=1$), which is commonly performed along the k_z axis with $k_x=k_y=0$, is identical to B_1^+ shimming as discussed in the introduction, i.e. a single RF pulse shape applied with channel-specific phase (20, 33, 34) or phase and magnitude (17, 19) modulation. (Throughout this work ‘ B_1^+ shimming’ denotes a 1-spoke excitation). All 2-spoke pulses designed in this study were based on a collection of 2-spoke locations sharing a point symmetry about $k_x=k_y=0$, with a radius $k_r/2\pi = (k_x^2 + k_y^2)^{0.5}/2\pi$ ranging from 0 to 10m^{-1} in steps of 1m^{-1} and an azimuthal angle ranging from 0° to 180° in 10° steps.

For all spoke excitations a same RF sub-pulse shape was used, consisting of a Hanning filtered SINC-shaped RF pulse with bandwidth-time-product of 4. In order to reduce

artifacts due to blood flow during the excitation we chose the RF duration to be as short as possible within the given SAR limits (see paragraph SAR computation and power monitoring); for the same reason, the gradients for the 2-spoke pulses were designed in a bipolar manner. RF pulse durations (without gradient ramps) were set to 1.6 ms for B_1^+ shimming excitation and to 0.8 ms for each sub-pulse of 2-spoke pulses, so that the total RF pulse duration for B_1^+ shimming and 2-spoke excitation was kept constant. Taking the gradient switching into account the 2-spoke RF pulse excitation is 0.4 ms longer.

In order to determine optimum tradeoffs between the residual error $\| |Ab| - |m_t| \|_w^2$ and the energy, L-curves were built by solving Eq. 3 while varying the value of λ , both for B_1^+ shimming and 2-spoke excitation. For each solution, the coefficient of variation (CV) of the transverse magnetization, defined as

$$CV = \frac{std_w(|Ab|)}{mean_w(|Ab|)} \quad [4]$$

, was calculated within the target region defined by w . Note that minimizing the quadratic residual error in equation 3 can provide solutions where $mean_w|Ab| < 1$; CV however is inherently normalized by the mean value of $|Ab|$ within the ROI defined by w . Additionally, the resulting total normalized energy E_n required to achieve $mean_w|Ab| = |m_t|$ with $|m_t| = 1$ was calculated by

$$E_n = \|b\|^2 \frac{|m_t|}{mean_w(|Ab|)} \cdot S \quad [5]$$

where the factor S takes into account the pulse duration for B_1^+ shimming ($S=1$) and 2-spoke pTX ($S=2$) (see Figure 3). Three different excitation settings were compared in-vivo:

- I. B_1^+ shimming
- II. 2-spoke solution yielding same CV as I (vertical line in Figure 3)
- III. 2-spoke solution yielding same total energy E_n as I (horizontal line in Figure 3)

For each solution, additionally to CV and E_n , the energy per individual transmit channel, E_i ($i=1 \dots K$), and the maximum value of the individual energies, $E_{i,max}$, were calculated with

$$E_{i,max} = \max(E_i). \quad [6]$$

All computations were performed in Matlab (The Mathworks, Nattick, MA, USA).

Hardware and experimental setting

Five volunteers, who signed a consent form approved by a local Institutional Review Board, were imaged on a 7 Tesla whole body MR system (Magnetom 7T, Siemens AG, Erlangen, Germany) equipped with a prototype 16-channel pTX system (Siemens AG, Erlangen, Germany). The 16-element RF coil consisted of two 8-transceiver arrays (35), positioned under the back and over the chest of the subject laying in supine position. Three electrodes were positioned on the chest skin for ECG recording. MR sequences provided by the vendor were modified to synchronize 16 transmit RF pulses simultaneously with ECG triggering. Calibrations were run to characterize potential time shifts between RF pulses and gradient waveforms; any measured temporal delay was incorporated into subsequent RF calculation.

SAR computation and power monitoring

Safe RF power operating levels were determined based on IEC (International Standard IEC 60601-2-33 2010) guidelines for SAR limits in normal and first level operation mode. For this purpose, electromagnetic simulation of the 16 channel transceiver body coil loaded with a whole body model was performed using a finite difference time domain solver (REMCORE, Pittsburgh, PA). A set of 160 B_1^+ shimming and 2-spoke solutions including those used for the experiments were computed for different orientations and corresponding local 10g average SAR values. The worst case of these solutions indicated that the IEC maximum local 10g average SAR limits over a 6 minute averaging window, i.e. 10W/kg in normal mode or 20W/kg in first level operation, would be reached when applying 1.49W or 2.97W, in normal mode and first level, respectively, on each of the 16 channels. In this study, a conservative setting was applied by limiting the maximum power per channel, continuously monitored by the vendor's multi-channel power monitoring system, to a maximum value of 2W on a 6 minutes averaging window. Following the IEC recommendation, the maximum limit was set to twice this value (i.e. 4W) on the 10s averaging window.

B_0 and B_1^+ mapping

B_0 and B_1^+ mapping is challenging due to heart motion and breathing motion. Thus, ECG triggering and a fast acquisition, i.e. within a 20 seconds breath-hold, is required to obtain clean maps. In addition, to reduce the impact of cardiac motion resulting from ventricle contraction during the systolic phase, both B_0 and B_1^+ maps were acquired with a 300 ms delay relative to the R-wave of the ECG.

ΔB_0 -maps, necessary to define matrix A in equation 3 for the 2-spoke RF pulse design, were obtained in a single slice for each view using a cardiac-triggered gradient echo sequence at two different TEs with following acquisition parameters: resolution= $2.8 \times 2.8 \times 5 \text{ mm}^3$, FOV= $450 \times 292 \times 5 \text{ mm}^3$, BW=801Hz/Px, TE₁/TE₂=2.3ms/3.3ms, TR₁/TR₂=4.3ms/5.3ms, corresponding to 550ms acquisition time per cardiac cycle. Four images were averaged for each TE. Note that the B_0 shimming routine of our system could not operate in pTX mode during this study, thus all experiments were acquired with the system tune-up shim.

To map B_1^+ of the individual channels we used a fast, multi-channel B_1^+ estimation method for transceiver arrays (36) that has demonstrated satisfactory B_1^+ map quality for multi-channel transmit methods at 7T (26, 33, 34, 37). Briefly, 16 complex small tip angle GRE images are acquired in single shot mode with cardiac gating, with a single TX channel active at a time, while all RX channels are used for reception, providing 256 images of distinct pairs of one transmit and one receive channels using the following acquisition parameters: resolution= $2.8 \times 2.8 \times 5 \text{ mm}^3$, FOV= $450 \times 292 \times 5 \text{ mm}^3$, TE/TR=2.6/4.7ms, BW=801Hz/Px. B_0 and B_1^+ calibration maps were obtained in all subjects in transversal and pseudo 4-chamber views; additional maps were collected in short axis view for 3 subjects. All acquisitions were performed under breath hold on exhale; neither partial Fourier nor Parallel Imaging was used. In two subjects an absolute flip angle map was estimated, using the AFI (38) sequence, under breathhold and without cardiac gating, with a 2-spoke excitation RF pulse.

Cardiac imaging protocol

CINE Gradient echo images were acquired in transversal, pseudo 4-chamber and short axis views using a modified cardiac triggered pTX sequence. For excitation setting I, the following acquisition parameters were used: resolution (non interpolated) = $2.3 \times 2.3 \times 5 \text{ mm}^3$, FOV= $450 \times 292 \times 5 \text{ mm}^3$, TE/TR=2.9/5.6ms, BW=554Hz/Px, segments=8. Reconstruction parameters included 2D interpolation (factor 2) and edge enhancement filter. All acquisitions were performed under breath hold on exhale. For setting II and III, as the total

RF pulse module duration is 0.4ms longer due to additional gradient switching compared to setting I, TE was about 0.2ms longer, and the time window that determines the temporal resolution increased from 41.7ms to 44.8ms. Up to 32 (B_1^+ shimming) or 30 (2-spokes) images were acquired per cardiac cycle. All other parameters were kept identical in setting II and III.

For excitation settings II and III, the voltage was adjusted to achieve the same flip-angle as obtained with setting I, which actually always resulted in lower associated power values, as will be detailed in results and discussion sections.

Results

Figure 1 shows the estimated magnitude (a) and relative phase (b) of the 16 B_1^+ maps, along a 4-chamber view, obtained in one representative subject (#3). Anterior and posterior coil elements are numbered 1–8 and 9–16, respectively, and the white ROI delimits the spatial target for B_1^+ shimming. It should be noted that these maps, well defined and free from significant artifact, were obtained without applying any spatial smoothing filter, which indirectly confirms the robustness of the fast, multi-channel mapping approach utilized in this study (36). One can also appreciate that the partial proton density bias, inherently observed with this fast estimation technique, has only a limited impact on the final B_1^+ maps. This residual bias, easier to identify when summing up the magnitude image of all individual B_1^+ maps as shown in Figure 2a, is assumed to be responsible for the apparent $|B_1^+|$ variations between lumen (higher) and myocardium (lower). In Figure 2b the corresponding ΔB_0 map, acquired with tune-up shim in the same slice, shows resonance offset frequency variations between –250 Hz and 150 Hz through the ROI. This cardiac-gated map, measured during diastole, was also free of any significant artifact, which is critical for the subsequent design of 2-spoke pTX RF. The necessity of including ΔB_0 into the 2-spoke RF pulse design is demonstrated in Figure 3c which shows simulations of the Bloch equation using a 2-spoke excitation for 3 different scenarios: i) no ΔB_0 off-resonance is present (left image); ii) off-resonance shown in Fig. 2b is present but not included in the pulse design (middle); iii) off-resonance in Fig. 2b is present and included in the design (right). As seen in the middle image, at the location of strong ΔB_0 with off-resonance frequencies beyond –200Hz, the flip angle is reduced by more than 50% without correction whereas including ΔB_0 in the pulse design effectively restitutes a satisfactory excitation profile.

In Figure 3a, the regularization term $\|b\|^2$, representing the total (non-normalized) RF energy, is plotted against the residual error of the excitation target $\| |Ab| - |m_t| \|_w^2$ for B_1^+ shimming and 2-spoke excitation, with λ values ranging over 8 orders of magnitude, resulting in both cases in an L-shaped curve. In Figure 3b, these data point have been normalized to obtain a same average target flip angle, with the resulting RF energy E_n (see Eq. 5) plotted against the CV value. One additional point (circle) in Figure 3b also shows the normalized result for the non-optimized starting B_1^+ phases with equal magnitude for the 16 TX channels, indicating a very inefficient B_1^+ shimming setting with poor homogeneity and high RF energy. These L-curves clearly show that with B_1^+ shimming as well as 2-spoke pTX excitation different tradeoffs can be chosen, setting a higher priority to reduce either CV or RF energy. In Figure 3b, the two curves converge on the right end, corresponding to the fact that, when a strong penalty is imposed upon RF energy, the optimal k-space locations for the 2 spokes are close, or equal, to the location of a single B_1^+ shimming spoke with $k_x=k_y=0$.

A major finding in this study is the significant advantage of using 2-spoke excitation: a same level of residual excitation errors (i.e. same CV) can be obtained with substantially less

energy (vertical gray line, setting II in Figure 3b) or, alternatively, for a same energy level, excitation errors can be substantially reduced (horizontal gray line, setting III).

This effect is quantitatively summarized in Figure 4 for all the volunteers where CV (Figure 4a), normalized total energy E_n (Figure 4b) and maximum energy per channel $E_{i,max}$ (Figure 4c) are displayed for all volunteers, settings and orientation. For example, in 4-chamber view, CV values vary between 11.1% and 15.2% for B_1^+ shimming and setting II (with identical values by design definition). However, the same CV and flip angle values were obtained, for all subjects, with a smaller total energy E_n for setting II which was reduced by a factor of 1.45 ± 0.19 (mean $\pm$ std) (see Figure 4b). Similar energy reductions were obtained in transversal (1.77 ± 0.25) and short axis views (1.35 ± 0.28). Alternatively, when a same E_n was preserved for B_1^+ shimming and setting III, the CV value was reduced on average by a factor 1.42 ± 0.17 compared to B_1^+ shimming in the four-chamber view and by a factor 1.47 ± 0.17 and 1.31 ± 0.30 for the transversal and short axis view, as shown in Figure 4c. Interestingly, in 2 out of 3 subjects, the energy reduction for the short axis view was less pronounced than for the four-chamber and transversal views, possibly the consequence of a smaller ROI where relatively low CV values could already be achieved with B_1^+ shimming. It was consistently observed that using 2-spoke pulses to reduce CV predominantly benefitted the posterior region of the heart where $|B_1^+|$ was typically low with B_1^+ shimming. This can be seen in a representative subject in Figure 5a showing transverse cardiac views acquired during systole and diastole for the 3 different settings, as well as corresponding simulated flip angle maps in Figure 5b. Using 2-spokes with setting III, the CV value was reduced from 16.6% to 12.1% and the contrast in the posterior heart section as well as the aorta could be recovered compared to B_1^+ shimming and setting II. Similar results are demonstrated in the 4 chamber view in Figs. 6a+b for B_1^+ shimming and setting III. Even though, by design, settings III and B_1^+ shimming had identical E_n , with CV much improved in setting III, the distribution of RF energy through individual channels was favorable for 2-spokes since $E_{i,max}$ was smaller in all orientations and all subjects, with an average reduction factor of 1.37 (see Figure 4). An even higher reduction factor of 1.81 on average was observed for setting II. The reduced $E_{i,max}$ value with 2-spokes has practical consequences, since a conservative maximum power limit was imposed on a *per-channel* basis.

The images shown in Figure 6a+b with B_1^+ shimming (left column) were acquired using a voltage close to the maximum allowed for the corresponding $E_{i,max}$. By keeping same flip angle and total energy but using setting III, not only CV value and signal intensity in the posterior heart areas clearly improved, but also $E_{i,max}$ was reduced by a factor >1.5 . Further gain in tissue contrast when using a higher flip angle could only be achieved with setting III where, thanks to the reduction in $E_{i,max}$, the nominal voltage could be increased by 22% (i.e. energy increase of 50%). This resulted in a clear contrast improvement between lumen and myocardium as shown in Figure 6a (right column), which is better visualized on a normalized line plot (Figure 6c) following the dashed line in Figure 6a after removing the estimated receive coil profiles from the data.

Especially when applying 2-spoke pTX pulse to cardiac imaging, a question arises about the validity of using a unique set of calibration maps (B_1^+ , B_0), measured during diastole, to calculate B_1^+ solutions to be applied through the entire cardiac cycle. However, as can be seen in Figure 7 showing 8 phases out of 32 (or 30 for 2-spoke) of the cardiac cycle (CINE acquisition), no significant differences in overall signal magnitude profile were observed at different phases of the cardiac cycle. For this particular comparison, B_1^+ shimming and 2-spoke excitation settings were chosen to have similar energy E_n , flip angle and CV values.

Discussion

Recently several groups have shown very promising cardiac images at 7T using multi transmit channel coils, driven either by a single amplifier with fixed phases between the TX elements or by multiple amplifiers in combination with B_1^+ (phase) shimming (4, 6–12). On the other hand, multi-spoke pTX pulses have proven to outperform B_1^+ shimming, providing better flip angle homogeneity and/or reduced RF energy in several applications, but has not yet been fully demonstrated in the human heart. Thus, the aim of this paper was to investigate the feasibility of using 2-spoke pTX excitation in cardiac imaging at 7T and to compare these results with B_1^+ shimming in the context of multiple outstanding methodological challenges.

Motion and blood flow are two major obstacles in cardiac pTX. Because of *respiratory* motion, fast calibration of B_1^+ or B_0 maps is typically restricted to single breath-hold modality (free breathing would require long calibration scans, incompatible with clinical settings). *Cardiac* motion requires ECG-based cardiac gating capability in MR sequences; in the case of pTX systems, this can present further hardware and software challenges as independent RF pulse design capability on each channel, with proper synchronization, is required on top of ECG triggering requirements. In the present work, these requirements necessitated to modify ECG-triggered MR sequences and to calibrate time shifts between RF pulses and gradient waveforms. With these modifications in place, robust multi-channel B_1^+ calibration maps, based on a small flip angle approach (36), could be acquired within one breath-hold (20s); cardiac diastole (300ms post ECG standard trigger) was found to provide the most efficient acquisition modality.

Blood flow may impact B_1^+ and B_0 map calibration, as well as final image quality when using 2-spoke pTX pulses. A short RF sub-pulses duration (800 μ s) was used in 2-spoke pulses to reduce the latter impact, while providing a reasonable tradeoff between duration and limiting SAR. Flow compensation was not included in that fast calibration scans to preserve short TE and short acquisition times that were of greater importance in the overall success of cardiac imaging at 7T. In all subjects, B_1^+ maps free of significant artifact were obtained to the cost of a partial bias in proton density, inherent to the chosen method, that did not prevent considerable gains to be obtained with B_1^+ shimming and 2-spoke pulses, even though non-biased B_1^+ maps may ultimately provide further quality improvement.

Mapping the absolute flip angle in the heart at 7T is still fairly challenging with current mapping protocols due to aforementioned issues, and obtaining such maps accurately is sometimes forgone to the benefit of more qualitative comparisons (39). Nevertheless, we were able to measure absolute flip angle maps using a 3D AFI acquisition in a breath-hold without cardiac triggering (AFI mapping is not compatible with adapting TR to each cardiac cycle), using the same 2-spoke excitation pulses as for the CINE acquisition. Based on these results we determined that the average flip angle in the heart when collecting images with our designed RF pulses was between 7° and 10°, consistent with our expectation. Naturally, the actual average flip angle may somewhat vary between subjects; however, using the same RF pulse design approach resulted in similar RF voltages in all subjects with similar contrast between myocardium and blood. In one particular case, taken here as an example, where the flip angle was 10°, the specific voltages required for a 90°, 0.5ms square pulse, varied between 40V and 352V through the 16 channels for B_1^+ shimming (voltages for 2-spoke excitation with same E_n 46V – 367V). Note, that the peak voltage for the 2-spoke excitation can be higher than for B_1^+ shimming although the maximum energy per channel $E_{i,max}$ is lower.

In 2 subjects, a slight reduction of signal amplitude was observed within the left ventricle in 4-chamber view with 2-spoke pulses compared to B_1^+ shimming, possibly the result of blood flow between sub-pulses (data not shown). Even in these cases, however, the overall image pattern closely matched the predicted excitation profile. Cardiac motion itself is another potential source of excitation pattern fluctuation; however, as shown in Figure 8, image quality was preserved throughout the cardiac cycle for both B_1^+ shimming and 2-spoke pTX.

The higher excitation fidelity obtained with 2-spoke pTX pulses, with improved image quality particularly in the posterior part of the heart, suggests that targets extending beyond the heart muscle, such as coronary arteries (13, 37) may greatly benefit from pTX methods. Alternatively, for a given CV value, 2-spoke excitation could be used to reduce total RF energy, with, in the current study, an average reduction factor of 1.77/1.45/1.35, for transversal/four chamber/short axis view, respectively. With 2-spoke pulses, a consistent reduction of $E_{i,max}$ was observed in all cases, which can have important consequences, especially when RF power limits are set on a conservative per-channel basis, since larger flip angle may be used with much improved tissue contrast.

It is of interest to note that several of the TX channels located on the bottom array do not contribute significantly to B_1^+ in the heart (see Figure 1), in agreement with previous simulation results (28). This suggests that with more optimal multi-channel coil arrangements, including coil element distribution along z-axis, B_1^+ methods could further enable cardiac imaging at 7T.

Conclusion

Despite major challenges such as motion and blood flow, 2-spoke pTX RF pulse excitation was successfully demonstrated for cardiac MR imaging at 7T. Compared to B_1^+ shimming, a better excitation fidelity or a reduced RF pulse energy could be achieved, opening the door to a broader range of applications, including uniform excitation for quantitative mapping approaches or coronary artery imaging at ultra-high field.

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Figure 1. B_1^+ magnitude (a) and phase (b) sensitivity profiles of the 16 channel transceiver coil in four-chamber view. The white contours delimit the target ROI.

Figure 2.

a) Sum of the B_1^+ magnitude profiles shown in Figure 1a in four-chamber view, with the same target ROI covering myocardium, lumen and aorta. b) Off-resonance frequency map, expressed in Hz, for same subject and same view as shown in a). c) Bloch simulations for 2-spoke RF pulses with/without including ΔB_0 in the RF pulse design and/or Bloch simulation. Left: ΔB_0 is assumed to be 0 for RF pulse design and for the Bloch simulation. Middle: ΔB_0 is assumed to be 0 for the RF pulse design but the actual ΔB_0 map shown in b) is incorporated in the Bloch simulation. Right: ΔB_0 map from b) is used in both, RF pulse design and Bloch simulation.

Figure 3.

a) L-curve plot of the two regularization terms $\|b\|^2$ and $\| |Ab| - |m_t| \|_w^2$ for B₁⁺ shimming and 2-spoke excitation with different values of the regularization parameter λ ranging over 8 orders of magnitude. B₀ variations were neglected for both plots. b) L-curve plot of normalized total energy E_n (Equ. 5) and CV (Equ. 4), derived from a). Except for CV, all quantities are given in arbitrary units. Compared to a) in plot b) both CV and E_n are normalized quantities taking the mean value of $|Ab|$ into account in order to achieve the target of $m_t=1$. Additionally, the RF pulse duration is taken into account for the calculation of E_n . The gray lines define points of same energy (horizontal line) and same homogeneity (vertical line) which were used to define the 3 experimental settings for this study.

Figure 4.

Quantitative summary of a) CV, b) E_n and c) $E_{i,max}$ for all volunteers and all views. In a) the CV for setting I and II is identical due to the study design whereas in b) E_n for setting I and III is kept the same. Short axis view data was only acquired in 3 out of 5 subjects.

Figure 5.

a) transversal view during systole (top row) and diastole (bottom row) obtained from CINE images acquired with excitation settings I-III. b) corresponding flip angle maps using simulations of the Bloch simulations for a target flip angle of 10° for the entire slice (top) and for the optimization ROI only (bottom). The flip angle is significantly reduced in the posterior region of the heart (see arrows) for settings I and II. Note the different color scales between both rows.

Figure 6.

a) pseudo 4-chamber view during systole (top row) and diastole (bottom row) obtained from CINE images acquired with excitation settings I and III only. Since $E_{i,max}$ was lower by a factor >1.5 with setting III (middle column) compare to setting I, the transmitter voltage/RF pulse energy could be increased by 22%/50% to obtain higher contrast values (right column). b) Corresponding flip angle maps for the entire slice (top) and for the optimization ROI only (bottom). Note the different color scales for the upper and lower row. c) Plots along the yellow dashed lines in a) to visualize the enhanced contrast by the increased transmitter voltage. The plots were corrected for the coil's receive profile.

Figure 7.

Cardiac images at 8 different phases of the cardiac cycle extracted from a 32-phase CINE acquisition, using either B_1^+ shimming (upper row) or 2-spoke (lower row) excitation, with corresponding flip angle maps obtained from Bloch simulations for a target flip angle of 10° in the rightmost column. Note the absence of noticeable difference in the overall profile of image signal magnitude between different phases of the cardiac cycle (from left to right) suggesting an absence of significant variation of excitation profiles. (B_1^+ shimming and 2-spoke settings were chosen so as to have similar energy E_n , flip angle and CV values).