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Seven-tesla time-of-flight angiography using a 16-channel parallel transmit system with power-constrained 3-dimensional spoke radiofrequency pulse design

Sebastian Schmitter, PhD^{1,*}, Xiaoping Wu, PhD¹, Edward J. Auerbach, PhD¹, Gregor Adriany, PhD¹, Josef Pfeuffer, PhD², Michael Hamm, PhD¹, Kamil Ugurbil¹, and Pierre-Francois Van de Moortele, MD, PhD¹

¹Center for Magnetic Resonance Research, University of Minnesota Medical School, 2021 6th St SE Minneapolis, MN 55455, USA

²Siemens Healthcare, Erlangen, Germany

Abstract

Objectives—Ultra high magnetic fields of 7 Tesla have proven to significantly enhance the contrast in time-of-flight (TOF) imaging, one of the most commonly used non-contrast enhanced MR angiography techniques. Compared to lower field strength, however, the required RF power is increased at 7 Tesla and the contrast obtained with a conventional head transmit RF coil is typically spatially heterogeneous.

In this work we address the contrast heterogeneity in multi-slab TOF acquisitions by optimizing the excitation flip angle homogeneity while constraining the RF power using 3D tailored RF pulses (“spokes”) with a 16 channel parallel transmission system and a 16 channel transceiver head coil.

Material and Methods—We investigate in simulations and in-vivo experiments flip angle homogeneity and angiogram quality with a same 3-slab TOF protocol for different excitations including 1-, 2- and 3-spoke parallel transmit RF pulses and compare the results with a circularly polarized (CP) phase setting similar to a birdcage excitation. B_1 and B_0 calibration maps were obtained in multiple slices and the RF pulse for each slab was designed based on 3 calibration slices located at the bottom/middle/top of each slab respectively. By design, all excitations were computed to generate the same total RF power for the same flip angle. In 8 subjects we quantify the excitation homogeneity and the distribution of the RF power to individual channels. In addition, we investigate the consequences of local flip angle variations at the junction between adjacent slabs as well as the impact of B_0 on image quality.

Results—The flip angle heterogeneity, expressed as the coefficient of variation, averaged over all volunteers and all slabs could be reduced from 29.4% for CP mode excitation to 14.1% for a 1-spoke excitation and to 7.3% for a 2-spoke excitations. A separate detailed analysis shows only a marginal improvement for 3-spoke compared to the 2-spoke excitation. The strong improvement

*Corresponding author, sschmitt@umn.edu.

The concepts and information presented in this paper are based on research and are not commercially available.

in flip angle homogeneity particularly impacted the junction between adjacent TOF slabs, where significant residual artifacts observed with 1-spoke excitation could be efficiently mitigated using a 2-spoke excitation with same RF power and same average flip angle. Even though the total RF power is maintained at the same level than in CP mode excitation, the energy distribution is fairly heterogeneous through the 16 transmit channels for 1- and 2-spoke excitation, with the highest energy for one channel being a factor of 2.4 (1-spoke) and 2.2 (2-spoke) higher than in CP mode. In vivo experiments demonstrate the necessity of including B_0 spatial variations during 2-spoke RF pulse design, in particular in areas with strong local susceptibility variations such as the lower frontal lobe.

Conclusion—Significant improvement in excitation fidelity leading to improved TOF contrast, particularly in the brain periphery, as well as smooth slab transitions can be achieved with 2-spoke excitation while maintaining the same excitation energy as in CP mode. These results suggest that expanding parallel transmit methods, including the use of multi-dimensional spatially selective excitation, will also be very beneficial for other techniques, such as perfusion imaging.

Introduction

The benefits of using ultra-high magnetic field (UHF) MR scanners for time of flight (TOF) angiography have been shown in studies performed at 7T by several groups¹⁻⁸. These benefits relate with higher spatial resolution, better vessel-to-background contrast as well as increased small vessel conspicuity and are mostly the result of longer T_1 relaxation time⁹ and of higher SNR¹⁰ inherently observed as the magnetic field increases. A main motivation for bringing TOF imaging to higher field, currently among the most commonly used MR angiography technique in the head, is to enable more accurate lesion depiction and better diagnostic capability, especially in small vascular lesions such as small aneurisms or cerebral stenosis. Furthermore, using UHF (7T and above) scanners for improving non-contrast-enhanced TOF imaging can be beneficial in patients with kidney conditions that prevent the use of exogenous contrast agents.

It is important, however, to note that TOF contrast is highly sensitive to any spatial nonuniformity of RF excitation. It relies on a short repetition time (TR) and a relatively high slab-selective excitation flip angle, typically about 15-40ms and 15-30° respectively, to saturate static tissue (dark background signal) while exciting fresh untagged blood flowing into the same volume (bright vessel signal). Also, TOF protocols often require high RF power levels, especially when magnetization transfer (MT) pulses are applied to improve background signal suppression and/or saturation pulses are applied on travelling slabs above (below) the excitation volume to suppress venous (arterial) vessel contributions. These are significant challenges at 7T because transmit B_1 fields (B_1^+) are inherently highly heterogeneous^{10,11} and because a larger amount of RF power converts into tissue heating at higher frequency, implying strict limitations to comply with SAR guidelines.

These challenges, however, can be addressed through pulse sequence optimization, RF pulse design, and with the use of multi-channel transmit RF technology¹²⁻¹⁴, as shown in two recent studies^{2,15}: 1) RF power reduction obtained by VERSE RF pulse design¹⁶ for saturation pulses and by sparse application of MT pulses allowed for using MT and SAT

pulses at 7T, resulting in improved vessel-to-contrast ratio (about +50% in the Circle of Willis) while keeping RF power within guideline limits, 2) using slab-specific B_1^+ shimming solutions in multi-slab 3D TOF, with a 16 channel B_1^+ shimming system, resulted in considerable improvement of B_1^+ homogeneity, thereby providing higher vessel-to-contrast homogeneity in brain angiograms with increased small vessel conspicuity especially in the brain periphery where excitation flip angle is typically much reduced at 7T when using conventional volume head RF coils.

In the current study, to further improve TOF quality at 7T (see disclaimer in Appendix 1), we bring into this MR imaging modality one more level of multi-transmit RF technology complexity by using multi-spoke RF pulses¹⁷⁻²¹ in combination with a 16-channel parallel transmit (pTX) system. In B_1^+ shimming^{15,22,23} (often labeled as ‘static’ B_1^+ shimming) one specific set of phase and magnitude (or phase only) RF input coefficients is applied to the transmit (TX) channels during an entire RF pulse, so that improving B_1^+ profile homogeneity is equivalent to improving *excitation* profile homogeneity. With multi-spoke pTX RF pulse design, however, the slab selective RF excitation pulse is split in a succession of sub-pulses, each of them played with a specific B_1^+ shimming set, separated by short gradient encoding blips, providing significantly larger degrees of freedom (DOF) to improve the final excitation profile. These additional DOFs give access to a larger range of RF pulse tradeoffs (increase B_1^+ homogeneity and/or reduce RF power requirement), that could not be obtained with B_1^+ shimming only²¹. To investigate the potential benefits of multi-spoke RF pulses, three RF excitation strategies will be compared in TOF angiography at 7T on a 16-channel pTX system: 1) Circularly Polarized (CP) excitation based on a set of RF *phases*, 2) 1-spoke excitation where B_1^+ shimming is performed with RF *phase and magnitude* modulation, and 3) 2-spoke excitation where different B_1^+ shimming sets in combination with gradient blips are applied on two subsequent RF sub-pulses.

Methods

Spoke RF pulse design: principle

The excitation homogeneity throughout a slab (or a slice) can be efficiently optimized using 3D tailored, or ‘spoke’, slab-selective RF pulses¹⁷⁻²¹. Here, an *S*-spoke RF excitation pulse consists of *S* serially applied filtered SINC-shaped sub-pulses together with slice selection gradients (with alternating polarity for subsequent sub-pulses). In between the sub-pulses, gradient blips are switched, with resulting excitation k-space trajectories illustrated in Figure 1 (in this study blip gradients are only applied along k_x - and k_y -axes). In the meanwhile, during each RF sub-pulse a specific set of 16 complex coefficients is applied on the 16 transmit RF channels. Designing the RF pulse consists in solving an optimization problem to find a complex vector $\hat{\mathbf{c}}$, containing $16 \times S$ complex coefficients, to approximately achieve, at the end of the RF excitation, the desired excitation pattern, defined here for the magnitude of the transverse magnetization. To this effect a magnitude-least-squares (MLS) optimization algorithm can be used²⁴, based on the spatial domain method presented by Grissom et al.²⁵:

$$\hat{b} = \underset{b}{\operatorname{argmin}} \left(\| |Ab| - |m| \|_w^2 + \lambda \|b\|^2 \right) \quad [1]$$

where the fidelity term $\| |Ab| - |m| \|_w^2$ quantifies the squared deviation of the excitation pattern $|Ab|$ from the target excitation pattern $|m|$, which is uniformly set to 1 in this work, and $\|b\|^2$ represents the total RF energy term, weighted by λ , the regularization parameter. Matrix A defines the excitation trajectory together with the timing of the excitation, the B_1^+ field of each coil element, the potential spatial deviations B_0 from the main field strength B_0 , and the spatial location of the excitation target. w denotes a binary mask defining the region of interest (ROI). Equation 1 can be rapidly solved by iteratively solving a modified version of the same equation where the *least-squares (LS)* difference $\|Ab - m\|_w^2$ is considered, instead of the MLS, while updating the phase of the target m at each iteration by multiplying the latter by the last residual phase of Ab , smoothed by a Gaussian kernel ²⁴.

Experimental setup and TOF protocol

All experiments were performed on a 7T system (MAGNETOM 7T, Siemens Healthcare AG, Erlangen, Germany) equipped with a prototype 16-channel pTX system using a 16-channel transceiver coil similar to ²⁶. Eight subjects who signed a consent form approved by a local Institutional Review Board were scanned in supine position. Multi-slab TOF acquisitions²⁷ with CP mode, 1-spoke and 2-spoke excitation were obtained in seven subjects. The eighth subject is analyzed separately in this study, since we were not able to achieve a suitable solution for the 1-spoke optimization as demonstrated in the results section. In the beginning of each study scout images and calibration scans were performed, including a 2-minutes low-resolution (1 mm isotropic) TOF scan to localize the Circle of Willis followed by acquisitions of B_1^+ and B_0 maps. RF excitation pulses were computed offline for each of the 3 TOF slabs and subsequently a TOF protocol with 0.5 mm isotropic resolution was acquired with 3 different RF excitations: i) CP mode; ii) 1-spoke and iii) 2-spoke RF pulses. All excitations were designed to have the same total energy E_n and the same target excitation flip angle of 20°. Imaging parameters were chosen as following: TR = 25ms; TE = 3.10ms (CP mode and 1-spoke), TE = 3.15ms (2-spoke), resolution = (0.5mm)³, 3 slabs with 24mm thickness each, 25% overlap, 25% slice oversampling, FOV = 220×172×60 mm³, acceleration: partial Fourier = 6/8 (along slice and phase axes), GRAPPA = 2.

TOF image reconstruction, including merging overlapping TOF slabs, was performed on the console using the vendor's original image reconstruction pipeline. Further TOF post-processed images, such as Maximum Intensity Projection (MIP) maps along different orientations and with different thicknesses, were performed in OsiriX (www.osirix-viewer.com).

CP mode excitation

A common challenge faced when using multi-channel TX head coils at 7T consists in obtaining a satisfactory starting set of transmit RF *phases* that prevents the formation of highly destructive B_1^+ interferences¹¹. Such interferences would otherwise result in signal

voids that could compromise subsequent required calibrations (B_1^+ and B_0 maps). This problem can be addressed by applying a set of phases aiming at reproducing a circularly polarized (CP) mode, resulting into an excitation pattern similar to that typically observed with a single TX channel volume coil, i.e. strong $|B_1^+|$ in the center of the brain with weaker $|B_1^+|$ in the periphery. In order to approximate this CP mode, $|B_1^+|$ maps were estimated in 4 equidistant slices covering 54mm along the head-foot direction using a fast calibration technique²⁸ that only requires a series of small-flip-angle GRE images acquired with one channel transmitting at a time. Based on the resulting complex B_1^+ maps, B_1^+ efficiency¹⁵ was optimized, altering RF *phase* but not RF *magnitude*, to match as closely as possible a 2D Gaussian-shaped spatial target as previously described^{15,29}. The corresponding $|B_1^+|$ solution with its set of RF phases will be referred to as “CP mode” throughout the study.

B_0 and B_1^+ mapping

After applying the CP mode set of RF phases, B_1^+ sensitivity maps were acquired for the 16 TX channels in 7 equidistant calibration slices, with 9mm inter-slice distance, covering 3 overlapping slabs that will later be utilized for TOF acquisition. Slabs and slices were positioned in such a way that each TOF slab comprised 3 calibration slices (upper edge, center, lower edge) with calibration slices #3 shared by slabs #2 and #3, and calibration slice #5 shared by slabs #1 and #2 (see Figure 2a). Channel-specific B_1^+ maps were obtained using a previously described hybrid approach³⁰ consisting of one absolute 3D B_1^+ map (AFI)³¹, acquired with all TX channels transmitting together in CP mode, merged with the complex relative B_1^+ maps of the 16 individual channels. These relative B_1^+ maps were obtained with a series of small-flip-angle 2D GRE acquisitions, at the 7 calibration slice locations, pulsing on a single TX channel at a time with all channels active for signal reception. The following parameters were used: i) AFI: TE/TR1/TR2 = 2.4ms/25ms/125ms; FOV = 256×200×252mm³; resolution=2×2×4.5mm³; nominal flip angle = 60°; parallel imaging: GRAPPA = 2. ii) GRE: TE/TR = 2.0ms/50ms, FOV = 256×200 mm²; in-plane resolution = 2mm × 2mm; 7 slices with thickness/distance = 3mm/9mm; nominal flip angle = 10°; no parallel imaging was applied. The reference voltage was set to 60V for all subjects, a value which consistently generated in CP mode an actual flip angle that did not exceed 80° in the center of the brain while matching approximately the nominal flip angle of 60° in the Circle of Willis.

B_0 mapping was performed on the same 7 slices using GRE images acquired at two TEs (TE₁=3.1ms, TE₂=3.6ms) with the following parameters: TR = 300ms, nominal flip angle = 30°, FOV = 256×200 mm², in-plane resolution = 2mm × 2mm; 3mm slice thickness.

Spoke RF Pulse design: practical implementation

An iterative optimization of the spoke RF pulse design was independently performed for each of the three 24-mm-thick TOF slabs overlapping each other by 6 mm as indicated in Figure 2a. The phase of the target m in Eq. 1 was initially uniformly set to zero unless otherwise specified. On each calibration slice an ROI defining a binary mask w_i ($i=1\dots7$) is drawn manually, covering cerebrum, cerebellum and brainstem. Note that the lowest slice of the bottom slice (calibration slice #1) had to be excluded from the ROI used for optimization because the local steep gradient of coil B_1^+ sensitivity along head-foot direction resulted in

very high RF power requirement to reach homogeneous excitation through the 3 slices of the slab; this bottom slice ROI was however included for all statistics characterizing the resulting excitation (mean flip angle and coefficient of variation). Within each slab, the corresponding 3 slice-wise masks w_i are then concatenated to form the ROI mask w used to define the system matrix A and the target vector m for Eq. 1:

$$A = \begin{bmatrix} A_1 \\ A_2 \\ A_3 \end{bmatrix}; m = \begin{bmatrix} m_1 \\ m_2 \\ m_3 \end{bmatrix} \quad [2]$$

Throughout this work, the RF pulse duration of each spoke is set to $s \text{ ms}/S$ (with $S=1$ or $S=2$), as shown in Figure 1, ensuring that the total RF duration, neglecting the gradient ramps and blips, is kept constant, independent of the number of spokes S . Each RF spoke consists of a Hanning-filtered SINC pulse with a constant bandwidth-time-product of 9.5. For each excitation the total normalized Energy E_n necessary to obtain a transverse excitation signal of $|m|=1$ is calculated according to Eq. 3:

$$E_n = \|b\|^2 \left(\frac{\text{mean}_w(|m|)}{\text{mean}_w(|Ab|)} \right)^2 S \quad [3]$$

In addition, the Energy E_k and the peak Voltage V_k applied to each TX channel $k=1, \dots, 16$ as well as the corresponding maximum $E_{k,max} = \max(E_k)$ and $V_{k,max} = \max(V_k)$ are noted. The excitation fidelity within the ROI w is quantified by the coefficient of variation (CV) defined as:

$$CV = \frac{\text{std}_w(|Ab|)}{\text{mean}_w(|Ab|)} \quad [4]$$

Spoke RF Pulse design: optimization

Before using the spoke RF pulse design in actual TOF experiments, the performance of 1-spoke, 2-spoke and also 3-spoke RF excitation pulses was first investigated in depth in one subject, based on B_1^+ calibration maps acquired as described above. In the case of 3-spoke pulses, each RF spoke duration was set to 670 ms, using the same RF spoke placements as for 2-spoke and an additional 3rd spoke placed at k-space center ($k_x=k_y=0$). The solutions for the 2- and 3-spoke excitation were calculated over a large range of excitation trajectories with corresponding polar coordinates (k_r, ϕ_k) defined as follows: a symmetric placement of the spokes in the k_x - k_y plane with respect to the k-space center was chosen with the distance $k_r/2\pi = (k_x^2 + k_y^2)^{0.5}/2\pi$ varying from 0 to 10 m^{-1} in steps of 1 m^{-1} and the angle ϕ_k ranging from 0° to 180° in 10° increments. For each trajectory, the value of λ in Eq.1 was varied over 10 orders of magnitude; for each of the latter, the MLS iterations to solve equation 1 were initiated with up to 32 different initial phase maps $\arg(m)$ defining the target m . The best out of these 32 solutions was then retained.

Based on this initial investigation it was determined that 3-spoke RF pulses would not be used in actual TOF experiments because only marginal gains were obtained with the chosen

3-spoke trajectory versus the 2-spoke solutions. In all subsequent TOF experiments, the 2-spoke RF pulse solutions were optimized using as subset of the spoke placements described above, always using a same initial target phase map ($\arg(m) = 0$), in order to limit the total calculation time. When needed, the optimization was repeated with a finer grid of λ values until a solution matching the desired energy target (described later) was achieved. In order to compare 1- and 2-spoke performance against CP mode, we chose, from the large stack of solutions found for each slab, the 1- and 2-spoke excitations that had the lowest CV values while providing the closest match between its total energy and E_n obtained in CP mode for the same average flip angle (see horizontal dotted line in Figure 3c). Another ROI, enclosing the larger part of the Circle of Willis, was drawn in the center slice (calibration slice #4, see also Figure 5b) in each dataset to calibrate the nominal flip angle for the CP mode, measured with the AFI sequence while using the same RF pulse applied in the TOF sequence. This calibration, played at a lower spatial resolution to preserve shorter scan duration, was used as a reference to determine the required voltage of all subsequent spoke RF pulses. All RF pulse computations were performed with custom software written in Matlab (The Mathworks, Nattick, MA, USA). Tikhonov regularization was implemented using the regularization tool package developed by Hansen³². In practice, RF pulse calculation times were about 90 seconds per slab for an array of 5 different λ values and all spoke trajectories of the 2-spoke solution, and less than 10 seconds for the 1-spoke solution.

Results

The tradeoff solutions between the energy term $\|b\|^2$ and the inhomogeneity term $\| |Ab| - |m| \|_w^2$ in Equation 1 are plotted in Figure 3a (through these solutions, λ values vary by about 10 orders of magnitude) for 1-spoke RF pulses with phase and magnitude modulation. Note that in this initial investigation, performed without collecting TOF data, B_0 variations were not considered. Both LS- and MLS-based optimizations were compared and, in both cases, the impact of different starting points for the target phase maps was further investigated, using either 1 (black and green curves) or 32 (red and blue curves) initial phase map(s) for the target magnetization m . In agreement with previous reports²⁴, a clear shift of the L-curve towards the origin was obtained when using MLS instead of LS. Interestingly, a step-like accident along the L-curves was observed for 1-spoke MLS optimization (green curve) when using a single starting phase map for m . We attributed this behavior, observed in several subjects, to the non-linear properties of MLS optimization. Increasing the number of starting phase sets for the target magnetization dramatically mitigated this effect, with the resulting curve remarkably closer to a traditional L-shape (blue curve in Figure 3a). A comparison between 1- and 2-spoke optimal solutions ($S=1$ and $S=2$, respectively), using 32 starting target phases, is plotted in Figure 3b (red and blue curve, $\varepsilon=1$), where each cross for a 2-spoke solution corresponds to a different k-space placement of the spokes and/or to a different λ value. The optimal 2-spoke solutions can be followed along the solid red line; the corner of this smooth L-shaped curved is clearly shifted to the left when compared to the 1-spoke optimal solutions. Note that for large λ values (right end of the curves), i.e. high regularization weight, the optimal 2-spoke solutions become indistinguishable from the 1-spoke curve; for those cases it was found that

the optimal 2-spoke placement also falls at the k -space center with $k_t=0$. In practice, the solutions indicated by the red and dark blue curves ($\varepsilon=1$) do not guarantee a constant flip angle over the plotted solutions, with an inherent tendency for the optimization to significantly reduce the mean flip angle when using high λ values (right end of the curves). Normalizing these results by $\varepsilon=\text{mean}(|A_b|)/\text{mean}(|m|)$ in order to always achieve the same mean flip angle results in an increase of total energy E_n (see equation 3), clearly identified as an offset of the normalized curves on the same plot (black and green curve), especially towards the right end. An easier way to capture the cost in RF energy required to achieve higher homogeneity is to directly express, for a given CV value, the required total Energy E_n for the different RF solutions to reach the same mean flip angle, as shown in Figure 3c. For comparison, the CV value corresponding to the CP mode was indicated (large black cross) on the same plot (with same energy and same mean flip angle). Starting from this black cross and following the horizontal dashed line to the left indicates, when crossing the blue curve, a significant reduction in CV, from 27.2% to 14.5%, obtained with a 1-spoke excitation when for the same E_n as in CP mode. Further to the left, when crossing the left edge of the red scatter plot, the 2-spoke RF pulse design provides further CV reduction, down to 4.6%, whereas only marginal CV improvement, down to 4.4%, is obtained with our 3 spokes pulse design when crossing the edge of the green scatter plot. In order to estimate how far the optimization could benefit from relaxing the constraint of having a symmetric spoke placement with respect to the k_z -axis, we compared two different optimizations for a given λ value that corresponds to solutions close to the L-curve corner. The first optimization used the same symmetric 2-spoke trajectory as before. The second optimization utilized the same locations but every possible combination of two spokes locations was tested. Compared to the optimal solution of the symmetric trajectory, the optimal asymmetric spokes solution can improve the energy by 9.1% and the CV by 9.3%. Thus, even with symmetrical spoke placement, the largest part of the expected benefit is already obtained.

As shown in Figure 4, the mean CV value $\pm$ std obtained for the TOF experiments averaged over the 7 subjects amounts to $29.4\% \pm 1.9\%$ for the CP mode. This value is reduced to $12.1\% \pm 1.7\%$ / $14.8\% \pm 3.3\%$ / $15.4\% \pm 4.4\%$ for the top/middle/bottom slab with a 1-spoke excitation and is reduced even further to $5.7\% \pm 1.2\%$ / $7.9\% \pm 1.2\%$ / $8.2\% \pm 1.9\%$ with the 2-spoke solutions. In order to allow for a direct comparison, the same total energy E_n and the same mean flip angle were preserved through all TOF acquisitions. As previously mentioned, B_0 maps were included in matrix A for subsequent 2-spoke RF pulse design when actual TOF data were collected. On the other hand, a single starting phase map for m was considered during TOF imaging sessions in order to shorten the total computation time. Beside the dramatic overall improvement in CV with 1- and 2-spoke pulses, Figure 4a also shows in both cases a consistent trend towards higher CV values for the bottom slab position. This could be explained by the fact that the bottom slab is located at the lower end of the RF transceiver coil elements, where a significant overall drop in $|B_1^+|$ sensitivity typically occurs along the head-foot direction, that could only be partially corrected by 1- or 2-spoke excitation. This mechanism would also be in agreement with the fairly constant CV ratio between the CV values obtained with 1-spoke divided by those obtained with 2-spoke

excitation, seen in Figure 4b. Through all subjects and all slabs, CV values are reduced by an average factor of 2.0 ± 0.5 for 2-spoke solutions relative to 1-spoke solutions.

Despite the fact that the 1-spoke and 2-spoke solutions shown here utilize the same total energy as the CP mode, voltage and energy distribution over the 16 channels is highly heterogeneous as demonstrated in Figure 4c based on a representative data set (top slab, subject 4). Here the top diagram displays the peak RF voltage per channel, normalized against the peak RF voltage in CP mode, for 1-spoke (black line) and for each sub-pulse of the 2-spoke excitation (two gray lines). The corresponding energy per channel is shown in the bottom diagram (averaged through spokes for 2-spoke excitation). It can be seen that the maximum peak RF voltage, through all channels and spokes, for the 2-spoke solution (here on channel 12) is higher than for the 1-spoke solution (here on channel 8); a similar observation was made through 20 out of the 21 slabs studied (7 subjects, 3 slabs each) with an average ratio of 1.28. Even though a practical consequence of using these solutions unaltered would be a need for higher maximum RF power capability with 2-spoke RF pulse design, it is important to note that, when taking the 2 spokes into account, the total energy per channel tends, on average, to be reduced with 2-spoke compared with 1-spoke excitation. A lower(equal)(higher) energy was required for the 2-spoke solution in 11(2)(8) out of the 21 imaged slabs, as summarized in Figure 4d, with an average ratio of 0.92 relative to the 1-spoke solution. The box plot in Figure 4d compares the maximum energy per channel between 1-spoke and 2-spoke for the 3 slabs in all volunteers. Although not statistically significant, a trend towards lower energy per channel is visible for the 2-spoke solution.

The effect of using an increasingly homogeneous excitation with CP mode, 1- and 2-spoke excitation on TOF angiograms can be appreciated in Figure 5a while the corresponding flip angle maps (resulting from simulations of the Bloch equation) are shown in Figure 5b. In general, spoke excitations are particularly beneficial in the brain periphery, as seen in transverse MIP images, where local B_1^+ defects in CP mode result in local losses in vessel visibility. Although 2-spoke and 1-spoke TOF results may, at first sight, look fairly similar in transverse MIP views, a closer examination indicates that significant vessel depiction improvement was obtained in the orbital branches of the anterior cerebral artery (yellow arrows) with 2-spoke excitation. Moreover, the coronal view shows locations of stronger vessel-to-tissue contrast with 2-spoke excitation compared to 1-spoke as indicated by the white arrows. This is consistent with regions where lower flip angles were expected as seen in the flip angle estimation maps for the 1-spoke in Figure 5b. Another consequence of residual flip angle variations when using slab-wise 1-spoke excitation takes the form of ‘venetian-blind’-like artifacts in coronal or sagittal views when the $|B_1^+|$ profiles of two adjacent slabs significantly differ at the junction between the latter, due to variations in background tissue suppression during TOF acquisition. Such artifacts can clearly be reduced when using 2-spoke excitation, which is further highlighted in Figure 6 showing, for the same subject, coronal views extracted from the original 3D TOF images (before MIP) as well as from 40mm thick MIP images when using a 1-spoke or a 2-spoke pulse. Note the high consistency between the reduction of the venetian blind effect (white arrows) and the improved spatial continuity between the flip angle maps obtained within the same calibration slice (used as top slice for middle slab as well as bottom slice for top slab, see red

boxes in Figure 5b) when applying the two corresponding RF pulses computed for the middle and top slabs during TOF acquisition (note that the colors scale was adjusted to emphasize the residual flip angle variations).

Another remarkable feature in 2-spoke excitation TOF acquisitions, seen in Figure 5 and further emphasized in Figure 7, consists in a substantial reduction of the subcutaneous fat signal compared with CP mode or with 1-spoke RF pulses. This effect was observed in all subjects. The range of the fat signal reduction varied from 1.64 to 2.83 with an average fat signal reduction of a factor 2.13 ± 0.45 , as measured in coronal 5mm thin MIP images in the posterior skull, close to the sagittal sinus. Since TE was chosen in this study to be as short as possible, the echo time for 1-spoke excitation is set to 3.11ms, which, at 7T, means that fat and water signals are almost in phase at TE and intra-voxel signal from both components add up constructively. Although the TE measured from the center of the 2-spoke RF pulse, i.e. in between the pulses, is fairly similar (3.15ms), the actual TE measured from the center of each individual sub-spoke amounts to 2.60ms and 3.70ms, respectively, which means that at 7T water and fat signals are close to 180° out of phase for each of these two TEs, resulting in destructive interferences between the corresponding signals. Although phase/water signal interferences are a very well-known, extensively exploited effect (a similar signal reduction can be achieved with 1-spoke pulses by simply adjusting TE, data not shown), it remains of significant interest to report that, with proper timing, water/fat cancellation can potentially be preserved in a straightforward manner in 2-spoke RF pulse design, independently of the sets of RF phases applied to each sub-pulse.

Achieving a nominal target flip angle is most challenging in the lowest slab, mostly a consequence of a rapid drop in $|B_1^+|$ sensitivity along the head-foot direction in this area, which typically results in higher RF energy demands to achieve a given excitation fidelity than in the middle or top slabs. In one of the subjects (s8) we were not able to obtain a suitable solution for 1-spoke excitation within the energy constraint defined by the CP mode taken as a reference: the corresponding excitation patterns always included some deleterious areas of very weak $|B_1^+|$ such as a dark central spot shown in Figure 8 (top row). Whereas an acceptable solution within the energy constraint could be found with a 2-spoke excitation design as shown in the bottom row of Figure 8, the only possible way to achieve an acceptable excitation pattern with 1-spoke was to allow for using 135% of the total energy defined in the CP mode (Figure 8 middle row).

An important requirement for in-vivo application of multi-spoke excitation at 7T is to address large frequency shifts due to B_0 modulations through space. As demonstrated in Figure 9a, frequency shifts of up to ~ 500 Hz are observed in the frontal area above the nasal cavity and in the lower temporal lobes. For CP mode and 1-spoke excitation, such off-resonance does not have a tangible effect on native TOF images or on the corresponding MIP based angiograms, as seen in Figure 9b. If a 2-spoke RF pulse excitation is calculated without including the B_0 distribution (i.e. without including B_0 maps to define matrix A in equation 1), however, strong artifacts occur with, in the frontal area, a visible crescent-shaped area void of signal in native TOF images (see Figure 9b, 3rd row, left image). Correspondingly, MIP based angiograms show disrupted vessel paths in the same location (same figure, right image). Including measured B_0 maps in the calculation, as done in this

study, dramatically reduces the size and extent of the corresponding artifact, as seen in the sagittal and transverse views (Figure 9b, 3rd and 4th rows, and Figure 9c, yellow arrow). These findings are consistent with Bloch simulations that are shown in Figure 9d at the level of calibration slice 2 and 3 for all four settings.

Discussion

2-spoke versus 1-spoke or B_1^+ shimming

In this work we experimentally demonstrate that improving excitation fidelity with 1-spoke and 2-spoke RF pulses on a 16-channel pTX system at 7T can provide substantial vessel depiction improvement in TOF cerebral angiography. Compared with a previous report on TOF at 7T with B_1^+ shimming¹⁵, the current study investigates the use of more advanced hardware resources as well as more advanced pulse design algorithms. With the multi-transmit-channel prototype console utilized in this study it was possible to modulate channel-wise RF *phase and magnitude* to determine B_1^+ shimming solutions, labeled 1-spoke through the study, instead of being limited to RF *phase modulation only* as in¹⁵. The same hardware resources further allowed applying 2-spoke RF pulses, known for their larger degrees of freedom to improve excitation fidelity²¹. The first main result of the current investigation is that, when imposing the same total RF energy and the same mean flip angle, 2-spoke excitation pulses optimized for a 24 mm thick slab can provide a substantial additional advantage over 1-spoke slab-specific excitation pulses, with a coefficient of variation reduced by almost one half within each of the three 24mm-thick slabs (in addition to the CV reduction by about one half obtained with 1-spoke excitation versus CP mode). Although all TOF angiograms obtained with 1-spoke or 2-spoke pulses showed dramatic improvements when compared with CP mode excitation, using 2-spoke pulses still brought tangible improvements over 1-spoke results. The 2-spoke excitations avoided residual areas of weak excitation, thereby preserving better local vessel-to-tissue contrast, and reduced inter-slab residual variations in excitation profile, thereby reducing venetian blind artifacts at the slab junctions. Furthermore, it is possible to preserve water and phase spins being out of phase with 2-spoke pulses to reduce subcutaneous fat signal.

RF power and RF pulse design

The importance of considering RF power requirement cannot be overstated when considering multi-channel RF tradeoffs. Indeed, as indicated in previous reports^{15,21} as well as in our current results (see Figure 3), it is doable to further improve homogeneity (i.e. reduce CV) up to a certain degree with 1-spoke or with *phase-only* B_1^+ shimming solutions, but this always comes at a certain cost in RF power that can become impractical because of either hardware limitations or patient safety guidelines, especially at 7T. Such limitations are exemplified when addressing venetian blind artifacts at the boundaries between adjacent slabs in TOF: we showed in a previous report that mitigating this artifact at 7T (using a similar coil as in the present study) with B_1^+ *phase* shimming only could require up to 3.8 times the power of the CP mode, which was used as a reference, for the same mean flip angle. Here we show (Figures 5 and 6) that such artifact can efficiently be addressed with 2-spoke excitation using the same total RF energy as in CP mode, whereas 1-spoke pulse failed at the same RF energy level. Another dire situation often encountered with B_1^+ phase

shimming or 1-spoke pulses is that some typical solutions obtained with MLS-based algorithms may consist in a relatively homogeneous $|B_1^+|$ distribution, except for one central area almost entirely devoid of $|B_1^+|$; such non-usable results are typically favored by the optimization when applying strong RF power constraints because these patterns (close to the so-called mode 0) are very efficient in terms of average flip angle per volt. In one subject, avoiding such non-usable solution (see Figure 8) with 1-spoke design would have required using 135 % the RF power of CP mode, whereas 2-spoke excitation easily converged onto excellent excitation homogeneity at no additional cost in RF power.

We also calculated the global SAR and local 10g SAR for the same sequence using the same optimization performed on an electromagnetic model of the used coil obtained with a finite difference time domain solver (REMGCOM, Pittsburgh, PA). The resulting global SAR values generated for the 20 degree flip angle amounted to 0.12 W/kg for the CP mode, 0.16/0.16/0.18 W/kg (bottom/middle/top slab) for the 1-spoke excitation and 0.14/0.13/0.15 W/kg for the 2-spoke RF pulse. Corresponding maximum local 10g SAR values amount to 0.6W/kg for the CP mode, 0.7/1.6/1.2 W/kg for the 1-spoke excitation and 0.6/0.6/0.7 W/kg for the 2-spoke excitation. The 2.3 times higher 1-spoke local SAR values compared to CP mode and 2-spoke excitation with same RF energy indicate that ultimately constraining local and global SAR rather than total RF energy is desirable. Such constraint, however, requires a more complex optimization that was not implemented at the time of the study.

Caveats with 2 spokes and B_0

Beside these advantages, some caveats have to be recognized with 2-spoke RF pulse design. The first issue concerns susceptibility-induced B_0 that can yield several hundred Hz of frequency offset and is inherently highly damageable in multi-spoke RF pulses if not corrected for. In spoke RF pulse design, however, this issue can be directly be addressed by collecting B_0 maps, a fairly fast and simple procedure, and integrate them in matrix A of Eq. 1. As demonstrated in Figure 9, this approach can dramatically reduce in size, and shift in space, areas of signal loss due to B_0 . At the same time, the bottom row of Figure 9b shows the persistence of a dark area after correction, even if of smaller size. This was attributed to the nature of very steep local 3D gradients of B_0 that are not expected to be fully compensated in our current implementation where RF spokes are played in 2 dimensions (k_x and k_y) and B_0 maps are only sparsely sampled along the z axis. Beside expanding B_0 sampling along z and playing spokes along k_z as well, other approaches could be merged with multi-spoke excitation to further mitigate this issue, such as using different time delays between transmit channels as proposed by Deng et al.³³.

Note that further investigations will be required to determine how much spokes that are played along k_x , k_y and k_z , instead of k_x and k_y only, may also help reducing RF power, especially at the bottom of the RF coil where rising the average flip angle in the lowest slice was fairly costly in RF power. Likewise, it remains to be seen whether including the k_z axis for spoke excursion may increase the benefit of using a 3rd spoke.

The basis of our investigation to compare different excitations was to set a constant total energy E_n for all excitations. In contrast to the CP mode, however, the 1- and 2-spoke optimization leads to a heterogeneous distribution of the energy over the different RF

channels, thus the energy $E_{k,max}$ required on the most demanding channel can be higher than the energy used for all channels in CP mode. Through all slabs and subjects, $E_{k,max}$ was found to be on average about 2.4 times higher with 1-spoke and about 2.2 times higher with 2-spoke solutions than in CP mode. It should be noted, however, that our pulse optimization only considers the *total* energy, not $E_{k,max}$ per channel, which could in principle be further controlled during RF pulse design. Furthermore, it should be reminded that on a 16-channel system each channel requires dramatically less power from each RF amplifier than the single channel of a conventional volume head coil, which translates in much smaller costs hardware-wise for a given relative increase of the maximum available peak power per channel.

Conclusion

In this study, significant advantages were demonstrated on cerebral TOF angiography quality in humans at 7T when using 2-spoke rather than 1-spoke RF pulse design on a 16-channel transmit system. Although most B_1^+ profile-related artifacts were better addressed with 2-spoke than with 1-spoke excitation, some challenges occur with 2-spoke pulses when addressing very strong local B_0 variations; existing solutions to address this issue were discussed. Other important methodological aspects of this study include the impact of spoke RF pulse design on RF power management. The positive results obtained in this study, using more advanced multi-transmit RF pulse design than previous reports relying on B_1^+ phase shimming, provide a strong incentive towards developing further new multi-channel RF transmission approaches for MRI-based perfusion and angiography in humans, such as multi-dimensional spatially selective RF excitation.

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Figure 1.

Excitation k-space trajectory (top) and corresponding scheme of the sequence's excitation part for 1- and 2-spoke excitation. The energy in k-space is deposited on the blue solid lines while the dotted blue lines denote the gradient blips for the 2-spoke excitation. The RF sub-pulse duration of the 2-spokes is set to 1ms, thus the total RF pulse duration is identical to the 1-spoke RF pulse.

Figure 2.

a) Locations of the 3 overlapping TOF slabs and the corresponding calibrations slices 1-7 used for B_1^+ and B_0 mapping. Slices #3 and #5 are shared for the optimization of two adjacent TOF slabs, respectively. b) transverse view of the sum of magnitude of the 16 B_1^+ maps including the ROIs w_i used for optimization. The ROI w_1 for calibration slice #1 was not used for optimization (see Discussion). c) B_1^+ magnitude maps of the individual TX channels for calibration slice #6

Figure 3.

a) L-curve plot demonstrating the tradeoff between the two regularization terms in equation 1, the energy term $\|b\|^2$ and the inhomogeneity term $\| |Ab| - |m| \|_w^2$. The black and red curves show the results for a least-squares optimization with 1 and 32 different target phases, respectively, while the green and blue curves plot the results for the MLS optimization. b) L-curve and scatter plots for the solution of 1- and 2-spoke MLS optimization using 32 different target phases without ($\epsilon=1$) and with ($\epsilon=\text{mean}(|Ab|)/\text{mean}(|m|)$) correction to obtain the same target flip angle for all solutions. The green and red

crosses indicate the individual solutions for the 2-spoke excitation having different excitation trajectories and/or λ values and the solid lines indicate the edge for the optimal solutions. c) L-curve and scatter plots showing the tradeoff between total (normalized) energy E_n and CV for 1-, 2- and 3-spoke excitation (blue, red and green) as well as for the CP mode (black cross). The horizontal line indicates solutions having the same total E_n as the CP mode, which was the design criteria for this study.

Figure 4.

a) Mean CV value and standard deviation from 7 subjects for the CP mode and for the individually optimized slabs for 1- and 2-spoke excitation. b) CV ratio, i.e. the CV values for the 1-spoke divided by the CV values for the 2-spoke excitation for each subject and each slab. The dashed lines show the average ratio value. c) Normalized voltage (top) and normalized energy E_k (bottom) for each of the 16 TX channels for 1- and 2-spoke excitation of the top slab in subject 4 normalized by the CP mode voltage (dashed line). d) Box plot for the maximum energy per channel ($E_{k,max}$) of all subjects split for the individual slab. The box indicates the median and the 25th and 75th percentile, crosses denote values (outliers) outside of $2.7 \cdot \sigma$ if a normal distribution is assumed and the whiskers extend to the extreme values which are not considered as outliers.

Figure 5.

a) Maximum intensity projections (MIP) for axial, coronal and sagittal views for CP mode, 1-spoke and 2-spoke excitation. b) Corresponding flip angle maps estimated by Bloch simulations for a nominal flip angle of 20°. The black ROIs denote the regions used for optimization and voltage adjustment. As shown for the 1-spoke and 2-spoke solutions, the slices #3 and #5 are used for the optimization of 2 adjacent TOF slabs respectively. The red boxes indicate the flip angle map locations used in Figure 6.

Figure 6.

Coronal native images and maximum intensity projections (MIP) for the same subject as in Figure 5 with 1-spoke (a) and 2-spoke (b) excitation. The right column shows flip angle maps estimated of calibration slice #5 which is used for the optimization of both, the top slab and the middle slab (compare red boxes in Figure 5). Note that the color scale of the flip angle, which is given in percent of the 20 degree target flip angle, is different compared to Figure 5.

Figure 7.

Transversal and sagittal views of native TOF images and corresponding 15-mm-thick MIP images for CP mode, 1-spoke and 2-spoke excitation. As indicated by the arrow, the fat signal is reduced for the 2-spoke excitation.

Figure 8.

Calculated flip angle map for the bottom slab in subject 8 for 1-spoke (top and middle row) and 2-spoke excitation (bottom row). In this subject a sub-optimal excitation showing a strong local flip angle void in the center of the slices (top row) could only be achieved for the 1-spoke excitation with the constraint of having the same total energy (100%) as the CP mode. A suitable solution, that successfully removed this local flip angle hole, was only found if the excitation energy was increased to 135% (middle row). In contrast, this effect was not observed for the 2-spoke excitation using 100% energy (bottom row).

Figure 9.

a) Off-resonance frequency maps acquired at the 7 calibration slice positions showing strong off-resonance in the lower frontal and the lower temporal lobe. b) Native images (left) and 15-mm-thick maximum intensity projections (right) in sagittal view with CP mode, 1-spoke and 2-spoke excitation. The 2 spoke calculation was performed without (3rd row) and with (4th row) including B_0 -induced off-resonances into the RF pulse design. This has a significant impact on the images and angiograms in the lower frontal lobe (yellow arrows). c) Corresponding transverse views for the 4 different excitation settings, showing a slice

located at calibration slice position 2 (see white box in subfigure a). d) Bloch simulations for the four different settings at the location of calibration slice position 2 and 3.