Frequency-adjustable magnetic field probes

Niklas Wehkamp | Philipp Rovedo | Elmar Fischer | Jürgen Hennig | Maxim Zaitsev

Faculty of Medicine, Department of Radiology, Medical Physics, Medical Center—University of Freiburg, Freiburg, Germany

Purpose: Nuclear Magnetic Resonance field probes provide exciting possibilities for enhancing MR image quality by allowing for calibration of k-space trajectories and/or dynamic measurement of local field changes. The purpose of this study is to design and build field probes, which are easier to manufacture and more flexible to use than existing probes.

Methods: A new manufacturing method is presented based on light-activated resin to encase the coil assembly and the 1H sample. This method allows for realizing field probes with tightly integrated orthogonal coils, whereby the local resonance frequency of protons can be adjusted during the MR experiment, by applying a DC current to the integrated B₀-field modification coil.

Results: The apparent field probe position in a gradient echo experiment was shifted within the field of view by changing its Larmor frequency using an integrated micro-coil with 5.5 windings. The measured frequency modulation induced by the B₀-field modification coil was 113 Hz/mA. The probe was tested with currents up to 100 mA. The DC current in the local field modification coil did not introduce visible artifacts in the MR images. Furthermore selective off-resonant excitation of the new field probes at 2 kHz above the main RF frequency was demonstrated. Gradient impulse response functions measured with a traditional and proposed probe show similar gradient imperfections.

Conclusions: The presented approach opens up new possibilities for concurrent field monitoring during MR experiments using standard RF capabilities of clinical scanners.

KEYWORDS
droplet formation, field monitoring, frequency adjustable, k-space trajectory, MRI, NMR probe

1 INTRODUCTION

NMR (Nuclear Magnetic Resonance) field probes offer great potential to improve MR (Magnetic Resonance) image quality. Such probes provide high resolution magnetic field measurements in the sub-ppm range as well as high bandwidths up to several MHz and thus are the superior choice for field monitoring during MR acquisitions over other sensing methods.¹
Field probes have been successfully used for MRI (Magnetic Resonance Imaging) system characterization, spatio-temporal field monitoring, real-time field control, trajectory mapping, artifact suppression in diffusion-weighted MRI, enhanced quantitative susceptibility mapping, and motion correction.

Hydrogen-based field probes provide the highest signal-to-noise ratio (SNR) but can interfere with the imaging process if concurrent field monitoring during the actual imaging experiment is desired. Therefore, fluorine, deuterium, or chemically shifted field probes have been proposed. A shortcoming of the magnetic field probes used so far for concurrent monitoring approaches is that they typically require additional transmit-receive electronics. This is because MR imaging experiments are conducted at a fixed resonance frequency, typically the Larmor frequency of protons, which is different to that of the magnetic field probes. The majority of commercial MRI scanners lack the capability of switching between different receive frequencies within the same pulse sequence. To circumvent this limitation, two approaches have been proposed: (a) to use completely separated transmit/receive electronics or (b) to mix the probe output with an additional carrier frequency to shift the probe signal into the proton receive band of the scanner. Both these approaches require a substantial technical effort. Another problem with concurrent field monitoring is that it is very difficult to suppress the signal coming from a field probe once it is excited, apart from allowing for signal relaxation to take place. Therefore, different MR imaging applications require the probe signal life and recovery times to be adjusted for optimal performance in the specific use case. In many circumstances it would be highly desirable to change the resonance frequency of the field probe in order not to interfere with concurrent MR imaging or to dephase probe signals without introducing strong global gradient pulses. Currently, different probes have to be used for each application mode in state-of-the-art field monitoring systems in order to fulfill the requirements of specific imaging applications.

This work presents a new design for hydrogen-based field probes, where the local Larmor frequency can be adjusted during a running MRI measurement. As the local Larmor frequency of each of the field probes can now be changed individually, numerous new opportunities concerning the mode of operation of the probes arise. For example, it allows to selectively promote or avoid probe excitation. In addition, it is possible to change the apparent position of the probe in frequency encoding direction and even shift the excited probe outside of the imaging field of view (FOV). These features enable the use of hydrogen-based field probes in cases where the commonly used fluorine-based field probes are impractical. Further possibilities arise, if the single field modification solenoid in the present prototype is replaced with a pair of coils, for example, to selectively apply “gradient” spoiling. In the current proof-of-concept study, we present the new manufacturing process and demonstrate the feasibility of the proposed prototype by measuring the frequency offset sensitivity and basic imaging experiments.

2 | METHODS

2.1 | Field probe assembly

Field probes as described rely on a glass capillary as a sample container. This glass capillary obstructs the positioning of a $B_0$-field modification coil in the vicinity of the field probe’s receive coil. In order to simplify the manufacturing process, the glass capillary was removed in the design presented here. For the current prototype we used the following bottom-up manufacturing approach depicted schematically in Figure 1: First, a droplet of
Ultraviolet (UV)-curing-glue (Blufixx MGS, BLUFIXX GmbH, Wesseling, Germany) was cured under a UV-lamp on a flexible Polystyrene surface. The cured droplet was detached from the Polystyrene surface and turned upside down. Two coils (receive—field modification coil) were placed on top of the droplet. The receive coil had an inner diameter of 1 mm, whereas the orthogonally oriented outer coil had an inner diameter of 1.8 mm. Both coils were wound from enameled copper wire (diameter: 0.2 mm). The receive and field modification coil were built with 3.5 and 5.5 turns, respectively. A second glue droplet was placed on top of the two coils but not cured. A droplet of water was injected with a pipette as depicted in Figure 2A,B. The 2.5-0.1 μl pipette (Eppendorf, Hamburg, Germany) was set to deposit 0.65 μl. Note that due to the higher viscosity of the glue compared to air, the deposited volume is smaller than the nominal volume. Subsequently, the second glue droplet was also optically cured. Additional layers of glue were applied in an iterative process of curing and adding glue until a spherical shape of the assembly was reached. Susceptibility matching was not done with the materials used for the presented proof of concept probes. Furthermore, the manufactured probes were not RF-shielded.

### 2.2 Simulation

The main goal of the simulation is to estimate the magnitude of the maximum frequency offset induced by the field modification coil. Thus, the influence of the wire thickness and the coupling with the receive coil was neglected for the presented simulations. The Biot-Savart law was used for computing the magnetic field \( \mathbf{B} \) at position \( \mathbf{r} \) in 3D-space generated by a steady current \( I \).

\[
\mathbf{B}(\mathbf{r}) = \frac{\mu_0}{4\pi} \int_C \frac{I d\mathbf{r}' \times \mathbf{r}' }{|\mathbf{r}'|^3}
\]

The line integral was evaluated over the conductor path \( C \) of our field modification coil discretized into pieces of length \( \Delta \ell \), along which the electric current flows. For the calculation the conductor path was divided into 360 parts per turn. For the six turns in the presented simulation, the path was divided into 2160 parts. It follows that the path length \( \Delta \ell \) of each element was set to \( 2\pi\text{radius}/360 \). The field and thus frequency offset was calculated on a grid of 100 × 100 × 100 within a volume of 50 × 50 × 50, 5 × 5 × 5 and 2 × 2 × 2 mm³, respectively. The field modification coil was simulated with an inner diameter of 2 mm, wire diameter of 0.2 mm and spacing between the conductor wires of 0.13 mm. The calculated field was multiplied by the gyromagnetic ratio of hydrogen and divided by the current yielding units Hz/mA. The matlab script for the simulation can be found at https://github.com/Nikker/solenoidSimulation.

### 2.3 Measurement methods

#### 2.3.1 Setup

The MRI experiments were performed on a 3 T MAGNETOM Prisma system with a 20-channel head coil (Siemens Healthcare, Erlangen, Germany). The system has a receiver bandwidth of ±250 kHz. The RF coil of the field probe was tuned and matched and connected to the scanner through a custom-built T/R switch and a low-noise preamplifier (Siemens Healthcare, Erlangen, Germany). The field probe’s RF coil was operated in receive mode. The field probe’s position and the excited FOV need to match in order to get signal from the field probe. The current in the field modification coil of the field probe was controlled with a power supply (HMP2030, Rohde & Schwarz, Germany) located in the control room of the MRI system. A 7 m long coaxial cable was used to connect the field modification coil to the power
supply. For the following imaging experiments, the field probe was fixed to the side of the water filled MRI phantom with tape to prevent it from moving. Figure 2C shows the frequency adjustable field probe on the MRI phantom inside the 20-channel head coil. The cylindrical MRI phantom had a diameter of 12 cm.

2.3.2 | Frequency shift

Frequency shifts of the new magnetic field probe were measured with respect to the current that was flowing through the $B_0$-field modification coil. For each current setting, the resonance peak’s maximum was evaluated in a 1H spectrum measured with the field probe’s RF coil. The MRI phantom was removed during this measurement in order to prevent interference with the recorded spectra.

2.3.3 | Gradient echo

The gradient echo (GRE) measurements were conducted to illustrate the shift in Larmor frequency of the new field probe. They were captured with a FOV of 160 mm, FOV phase of 100 %, a resolution of 250 × 250 pixel, slice thickness of 20 mm, TR of 100 ms, TE of 6.45 ms and a bandwidth of 100 Hz per pixel with a read oversampling factor of 2. The signal was combined from all available channels of the head coil and the field probes RF channel. The frequency encoding of the GRE measurement was set to the horizontal direction of the image. Thus, changes in the resonance frequency of an object in the image should result in an apparent change of position along this direction.

2.3.4 | Phase images

Phase images were reconstructed from GRE measurements with a FOV of 220 mm, FOV phase of 100%, a resolution of 250 × 250 pixel, slice thickness of 20 mm, TR of 50 ms, TE of 3.71 ms and a bandwidth of 260 Hz per pixel. The phase image acquired through the RF channel connected to the field probe is shown in the results section. The frequency encoding of the GRE measurement was again set to the horizontal direction of the image. The field probe was mounted at a close distance of 15 mm, to investigate if the $B_0$-field modification coil distorts the phase image of the MRI phantom. Then, two images were acquired with a modification current set to 0 mA and 100 mA, respectively. Phase images are particularly affected by external signal sources. Here, they are used to make areas where artifacts could occur visible. In order to visualize possible deviations between the two phase images, they were subtracted from each other.

2.3.5 | Selective off-resonant excitation

We relied on the pulseq framework\(^2\) to implement a custom GRE sequence for off-resonant excitation. Matlab environment (MathWorks, Natick, Ma, USA) was used both to program pulse sequences and analyze the recorded data. Trigger signals were added to the pulseq framework in order to control the $B_0$-field modification coil. The scanner supplied the trigger signals as optical output, controlling a custom built trigger box that switched the current provided by the power supply on and off. A solenoid driver IC (ULN2003B, Texas Instruments, USA) was used as a switch. The power supply was set to a current of 20 mA at a max. voltage of 20 V. For elimination of the currents on the shield of the 7 m coaxial cable a choke balun was inserted at the end of the cable entering the bore of the MRI scanner. In order to suppress high frequency signal from the control room a low pass filter was introduced at the feed-through panel from the control room to the scanner room.

The GRE sequence was extended with a Gaussian pulse for field probe excitation to demonstrate the off-resonant excitation of the new probe. The Gaussian pulse for probe excitation was set to the shifted resonance frequency of the field probe (ie, the frequency of the probe when the B-field modification current is on). For the here presented measurement the frequency was shifted by +2265 Hz compared to the center frequency of the MRI system. The frequency encoding direction for this set of measurements was in Z direction of the laboratory frame. The bandwidth of the main RF pulse and the Gaussian pulse was set to 1 kHz. The pulseq Matlab source files to generate the sequence with detailed sequence information can be found in the Supporting Information. Three pulse sequences were used to demonstrate the selective off-resonant excitation of the new field probe. The first sequence was implemented without main pulse and with the trigger turned on during the Gaussian pulse. In the second sequence the trigger was only turned on during the main pulse and turned off during the Gaussian pulse. The third sequence had the trigger turned on for the duration of both the main and the Gaussian pulse. Corresponding schematic sequence diagrams are depicted in Figure 7.

2.3.6 | Free induction decay (FID)

FIDs were recorded both after the excitation at the main RF frequency and after off-resonant excitation, respectively. Again, an existing pulseq sequence\(^2\) was modified to allow for switching a frequency offset current during measurements excitation. A 90° block-pulse with a duration of 10 ms was used for excitation. Calibration of the 90° pulse was done by varying the excitation voltage and calculating the 90° pulse after identifying the 180° excitation. The frequency offset was set to 0 during
the 0 mA and to 2265 Hz during the 20 mA measurement. A delay of 2 ms followed the excitation to ensure complete coil ringdown before acquisition in the receive coil of the probe. A delay of 2 ms is longer than necessary, but was chosen to be sure to exclude all influence. For signal acquisition the ADC duration was set to 500 ms with 1000 sample points followed by a second delay of 10 s to ensure full relaxation before the next measurement. Ten repetitions were recorded. The FID was recorded from the receive channel of the field probe only. The MRI phantom was removed during these measurements in order to prevent interference with background signals in the recorded spectra. The pulseseq source files can be found in the Supporting Information.

2.3.7 Z-gradient characterization by impulse response measurements

The $B_0$-modification coil is oriented along $B_0$ and thus along the $Z$-gradient direction. Therefore, especially $Z$-gradient switching is expected to induce currents and/or mechanical vibrations in the $B_0$-field modification coil. This may potentially translate into instabilities in the probe readout. The gradient impulse response function (GIRF) was determined to investigate if the response function of a probe with $B_0$-field modification coil shows a response different from a reference field probe with a receive coil only. The reference probe was assembled according to Section 2.1 disregarding the $B_0$-field modification coil. This may potentially translate into instabilities in the probe readout.

Knowledge of the GIRF could form the basis of advanced field probe coil characterization by impulse response measurements as described in reference \(^3\) to obtain the GIRF. The measured GIRFs were used to predict the field responses of an additional reference measurement. This measurement was recorded with the reference field probe that has only a single receive coil.

3 RESULTS

3.1 Simulation

Simulations have been performed for an idealized coil with six windings and neglecting the leads to and from the coil. The simulated frequency offset per current in Hz/mA induced by a current flowing through the $B_0$-field modification coil is below 0.01 Hz/mA for distances larger than 26 mm and even below 0.0025 Hz/mA for distances larger than 45 mm. Figure 3 illustrates the frequency offset with contour lines. The first excerpt in the middle of the figure shows how the magnitude of the B-field modification increases according to the Biot-Savart law to the power of 2 in proximity to the coil. At a distance of 4 mm, the frequency offset per 1 mA current is 2.5 Hz/mA. The second excerpt on the right illustrates the frequency offset inside the $B_0$-field modification coil. The measured frequency offset per current in Hz/mA is about 115 Hz/mA. The contour lines in the receive-coil show that the frequency offset per current will vary between 100-120 Hz/mA inside the receive-coil.

3.2 Frequency shift

Measured frequency shifts are depicted in Figure 4. The frequency and thus the coil’s field shows a linear characteristic in the measured range from 0 to 100 mA with a slope of 113 Hz/mA. At a current of 100 mA, the measured frequency offset is 11 kHz.
3.3 | Gradient echo (GRE)

Figure 5 illustrates the effects of the shifted frequency of the field probe in the conducted GRE measurement. The numbers in the lower right corner indicate the amount of current that was flowing through the \( B_0 \)-field modification coil during the measurement. A current of \( I = -20 \) mA shifts the field probe’s apparent position to the left (in frequency encoding direction) toward the top of the MRI phantom. Inverting the current \( (I = 20 \) mA), the frequency shift appears to move the probe toward the right of its actual position which is displayed in the \( I = 0 \) mA case. This clearly shows that the apparent position of the field probes can be manipulated in an MRI image by running a current through the \( B_0 \)-field modification coil. The broadening of the probe appearance in the image in frequency encoding direction is due to field inhomogeneity of the \( B_0 \)-field modification coil, which leads to spectral broadening with increasing current.

3.4 | Phase images

The phase image displayed on the left in Figure 6 was recorded with no current (0 mA) running through the \( B_0 \)-field modification coil. A cross-shape artifact is clearly visible in the upper right quadrant of the image. The phase image in the middle of Figure 6 was recorded with a current of 100 mA in the \( B_0 \)-field modification coil. Here, the cross-shaped artifact is not visible anymore, due to the current dependent apparent shift of the field probe. The difference of these two phase images depicted on the right in Figure 6, shows the cross-shape artifact without the signal from the MRI phantom. The interesting aspect here is that no visible artifacts in the MRI phantom were recorded. Especially the red circled area in the difference of the two recorded phase images demonstrates that the \( B_0 \)-field modification coil does not introduce artifacts at a distance of 15 mm.

3.5 | Selective off-resonant excitation

The possible off-resonant excitation of the field probe is demonstrated by the following three experiments. The GRE image in Figure 7 shows a bright dot at the location of the field probe. As the sequence diagram illustrates, the field probe had been excited by a frequency shifted Gaussian pulse and concurrently resonance frequency shifted with the
B-field modification current. In order to obtain substantial signal from the probe, the excitation frequency has to be closely matched to the frequency shift induced by the $B_0$-field modification coil of the field probe. This demonstrates the possibility to excite the probe off-resonance and acquire the signal at the main resonance frequency of the scanner. It is as well possible to hide the field probe from the excitation pulses as shown in Figure 7B. Here, the MRI phantom was excited with a sinc pulse at the main RF frequency of the scanner, while the B-field modification current was turned on. The second Gaussian pulse was again frequency-shifted by 2265 Hz, corresponding to the resonance frequency of the field probe when the B-field modification current is on. The image in Figure 7B only shows the image without any signal from the field probe. This demonstrates the possibility to selectively hide the field probe from both excitation pulses. In practical implementations, the off-resonant Gaussian pulse could of course be avoided. The signal was recorded with the RF channel of the field probe only, which explains the intensity gradient visible in the MRI phantom.

Figure 7C shows the case, when probe and MRI phantom were excited consecutively by two independent RF pulses while the signal was recorded simultaneously at the main frequency of the MR system. This shows that MRI phantom and field probe can be excited independently for example, with separate optimized flip angles.

### 3.6 Free induction decay

FIDs and the corresponding spectra at a modification current of 0 mA and 20 mA are depicted in Figure 8. At 0 mA B-field modification current, the $T_2^*$ of the field probe was 17.1 ms. This is considerably shorter than the $T_2^*$ of 100 ms of susceptibility matched probes in the literature. The materials in the probes presented here were not susceptibility matched, which
explains the relatively short $T_2^*$ at 0 mA. Measured FIDs from the probe at 20 mA modification current and off-resonant excitation showed a shorter $T_2^*$ of 9.3 ms at a frequency shift of 2.2 kHz. The reduction in $T_2^*$ is probably caused by B-field inhomogeneities introduced by the $B_0$-field modification coil. The homogeneity of the relatively short solenoid coil used in the present study was not optimal. It can potentially be improved by additional wire windings and the optimization of the winding pitch. An alternative approach to improve the relative field homogeneity across the sample volume at the cost of SNR is to reduce the water droplet size.

### 3.7 | Z-gradient characterization by impulse response measurements

The GIRFs of the Z-gradient are depicted in Figure 9A. All GIRFs show a similar behavior for frequencies below 12 kHz. The GIRFs do not exhibit any spikes or strong deviation compared to the reference measurement. With increasing frequency above 12 kHz, both differences between the GIRFs and the variance within each single GIRF increase. Furthermore, turning on the $B_0$-modification current did not change the response significantly, which suggests that the current source in use is stable enough to support even off-resonant field measurements.

Figure 9B depicts a nominal reference gradient trapezoid with the corresponding measured and GIRF-based predicted gradient waveforms. The measurement is in close agreement with the predicted gradient response. This becomes even more evident in the excerpt depicted in Figure 9C showing the overshoot of the Z-gradient after ramp up. This illustrates that the determined GIRF can be used to correct for systematic deviations of the gradient response from the nominal trajectory.

### 4 | DISCUSSION AND CONCLUSION

NMR field probes that allow for calibration of k-space trajectories and/or dynamic measurement of local field changes
have already proven useful in a number of MRI experiments. However, their wide application is limited by the necessity of separating their signals from the primary nuclear induction, as well as controlling their life time. Available probes operate at a single resonance frequency with the life time defined during manufacturing by paramagnetic doping to match typical intended applications. For concurrent monitoring, field probe signals are typically separated in the frequency domain using other nuclei (e.g., fluorine), which requires additional RF electronics.8,9,13,15

FIGURE 8  Top: FID of the field probe with a modification current of 0 mA (left) and 20 mA (right). Bottom: The two corresponding spectra with peaks at −22 Hz and 2245 Hz, respectively. The $T_2^*$ was 17.1 ms at 0 mA and was reduced to 9.3 ms at 20 mA modification current. The reduction of $T_2^*$ is also visible in the broadening of the peak at 2245 Hz

FIGURE 9  A, Magnitudes of the measured gradient impulse response functions (GIRF). A reference GIRF was measured with a field probe that has only a receive coil (1 coil). The GIRFs of the new field probe with a $B_0$-field modification coil (2 coils) and with the same probe and the modification current of 5 mA turned on (2 coils, 5 mA) show a similar behavior for frequencies below 12 kHz. At frequencies above 12 kHz, both differences between the GIRFs and the variance within each single GIRF increase. B, GIRF-based predictions of Z-gradient trapezoid compared with nominal and measured gradient waveform. C, Excerpt of (B) were all GIRFs predict a slight overshoot of the Z-gradient, which can be observed in the reference measurement as well.
Here, a new field probe concept is presented and demonstrated in several proof-of-concept experiments. Our proposed field probe features a second coil. This coil can be used to modify the B₀ field in the MR-sensitive region of the probe. The additional coil allows one to change the resonance frequency of our proposed field probe by running an electric current through it. To enable a two-coil configuration with both coils producing orthogonal magnetic fields (B₁ and B₀-field modification, respectively) the glass capillary known in the previous field probe designs was eliminated. This was achieved by a new manufacturing process based on the UV light-activated resin.

The design of the field probes presented here has the additional advantage that the number of different materials used during the manufacturing is reduced, which will simplify the susceptibility matching of the relevant components in future. Susceptibility matching was not performed during the manufacturing of the probes presented here, which can explain the relatively short signal life time of 17 ms that was observed. However, as presented in the Supporting Information section, the susceptibility of the light-activated resin used in this study can be altered with previously published methods.

The manufacturing of susceptibility matched prototypes is planned in near future.

Another critical aspect is the reduction in T₂* from 17.1 ms to 9.3 ms when a B₀-field modification current of 20 mA was applied to the probe. This shortening is probably caused by the inhomogeneity of the B-field of the employed B₀-field modification coil. The simulations presented in Figure 3 already indicate that the homogeneity of the relatively short solenoid coil used in this proof-of-concept work was certainly not optimal. The reduction in T₂* could limit the possible measurement scenarios. Thus the homogeneity of the B-field modification coil needs to be improved in order to increase application potential and allow for measurement scenarios requiring longer T₂*. The homogeneity can potentially be improved by adding more turns and optimizing winding pitch. Another approach to improve the homogeneity of the field over the sample volume is a reduction of the droplet size. Both approaches will be tested in future.

The results measured with the proof-of-concept field probe show a linear dependency of 113 Hz/mA between the applied field modification current and the resulting frequency response. The acquired GRE images demonstrate that it is possible to change the apparent position of the field probe in the obtained image. The corresponding phase images show clearly that there is no detectable effect of the local B₀-field modification coil on the phase images in the vicinity of the probe even with the modification current set to 100 mA when the probe signal is outside the FOV. The selective off-resonant excitation experiments illustrate potential use cases of the frequency-adjustable magnetic field probes.

Changing the resonance frequency of magnetic field probes will allow for more flexible experiment design, because the probes can now be turned “on” and “off.” For example changing the probe frequency during the signal excitation allows for selective excitation. During signal readout, this ability can be used to remove the probe signal from the ADC bandwidth or selectively dephase the probe signal (if a local gradient coil is used) to allow for faster probe recycling. Furthermore, concurrent measurements can be conducted at a shifted frequency to avoid interference with other MR measurements. The flexibility in the way the probe can be deployed has the advantage that it allows for permanent integration of the probes with current RF coils or parts of the MRI system.

The frequency shift allows one to put the field probe signal outside of the maximum imaging signal bandwidth defined by the object dimensions and the readout gradient, potentially allowing the field probe signal to be combined with a surface coil signal and fed into the same receiver. This would provide further hardware simplification by reducing the number of receive channels needed to support a field probe array. The probes presented here can, therefore, utilize the same receiver hardware that is used for the main MR measurement. Apart from the system costs, this can further facilitate synchronization of the measurements since the internal clock and the processing pipeline of both analog and digital signals is the same. The latter can be another important advantage compared to current fluo-
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DATA AVAILABILITY STATEMENT

The code that supports the findings of this study is available in https://github.com/Nikbe rt/solenoidSimulation. The pulseseq Matlab source files to generate the sequences can be found in the Supporting Information.

ORCID

Niklas Wehkamp https://orcid.org/0000-0003-3075-7904
Philipp Rovedo https://orcid.org/0000-0003-3836-2202
Elmar Fischer https://orcid.org/0000-0003-0412-0379
Jürgen Hennig https://orcid.org/0000-0002-2273-3497
Maxim Zaitsev https://orcid.org/0000-0001-7530-1228

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SUPPORTING INFORMATION

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

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