

A Purpose-Built Neck Coil for Black-Blood DANTE-Prepared Carotid Artery Imaging at 7 Tesla

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ABSTRACT

Atherosclerotic plaques in the bifurcation of the carotid arteries can pose a significant health risk due to possible plaque rupture and subsequent stroke. The assessment of plaques, and evaluation of the risk they pose, can be performed with Black-Blood (BB) vessel wall magnetic resonance imaging. However, resolution at standard clinical field strengths (up to 3 Tesla) is limited, hampering reliable assessment and diagnosis. The aim of this study was to investigate the benefits of 7 Tesla MRI using a BB application that has been successful at clinical field strengths. Therefore, for BB imaging, each sequence was preceded with 'Delay Alternating with Nutation for Tailored Excitation' (DANTE) preparation pulses for blood signal suppression. A coil comprising a 4-channel Tx array was designed and built to provide the required excitation coverage for the DANTE train; and a 4-channel Rx array was constructed to target the carotid bifurcation. Human and phantom results showed satisfactory blood suppression and comparable SNR and CNR to 3T, therefore demonstrating the feasibility of the application at 7 Tesla. However, the imposed SAR restrictions led to long scan times and subsequent motion artifacts. Thus, more accurate local SAR supervision schemes are required which could lead to a further improvement of BB DANTE vessel wall imaging at 7 Tesla.

Keywords: radiofrequency coil; ultra-high field; 7 Tesla; neck imaging

1 INTRODUCTION

Stroke, due to atherosclerotic emboli arising from the carotid arteries, is a subject of considerable clinical interest, since rupturing of plaques can cause irreversible brain damage and possibly death. In particular, unstable soft plaques are often considered vulnerable, posing a high rupture risk. Furthermore, the bifurcations of the carotid arteries to the internal and external carotid arteries are locations of high risk with respect to such ruptures due to increased shear stress [1,2].

As far as diagnosis of atherosclerosis is concerned, arterial imaging using a combination of clinical imaging modalities is the most accepted method. The imaging protocol often comprises ultrasonography [3,4], X-ray computer tomography (CT) or use of magnetic resonance imaging, Black-Blood (BB) imaging and/or MR angiography (MRA) [5,6,7]. The assessment of the plaque vulnerability in the carotid arteries with MRI has shown clinical prospect [8,9,10]. However, better definition is desirable for more accurate and reliable diagnosis of the plaque, and it is expected that higher quality images could be generated at ultra-high field (UHF) for this application [11,12].

In this study, specially-designed blood suppression preparation pulse trains were used before the read-out segments. The DANTE preparation module [13] comprises non-selective low flip angle pulse trains interspersed with gradient pulses with a short inter-pulse spacing. This results in preservation of the longitudinal magnetization of static tissue and attenuation of the signal coming from flowing spins, due to high order phase spoiling in the transverse plane that transfers to crushing of the longitudinal magnetization. A useful property of DANTE is that the signal suppression is largely insensitive to the velocity profile, which varies within the large cross section of the carotid arteries.

Body coils are impractical at 7 Tesla and techniques such as RF shimming with transmit arrays gaining popularity for addressing B_1 inhomogeneity [14,15,16,17]. Another issue is RF safety [18,19,20,21,22]. Although safety was considered in this study, the main topic is the design and construction of a suitable coil for DANTE BB imaging at 7 Tesla, and its preliminary results.

Studies of vessel wall imaging at 7 Tesla are relatively limited, however in [23] a comparison between 3 and 7 Tesla was made using double inversion recovery for blood suppression, showing improved images at 7 Tesla. In [24] an 8-channel Tx/Rx neck coil was constructed for BB carotid vessel wall imaging at 7 Tesla using a TSE sequence. In [25] carotid artery imaging with a travelling wave coil was presented, which could improve the field distribution. There is also a large number of papers in the literature describing optimized RF coil design at high field strength [26,27,28,29,30,31,32,33].

The overall goals of this study were to: (i) design and construct an RF coil suitable for carotid artery imaging at 7T; (ii) use the coil to perform black blood imaging with DANTE preparation pulses; and (iii) assess the feasibility of the coil for vessel wall imaging and determine the necessary next steps.

2 METHODS

Usually, MRI coils have a birdcage transmitter and multiple receiver loops surrounding the volume. For this application, however, there is a requirement for the transmit field to remain effective outside the ROI. This practically means that the volume transmitter needs to extend beyond the area covered by the receive array. Therefore, two separate arrays were built, one for Tx and one for Rx, and their design and construction is described below.

2.1 Transmit Array Design

The design process for the transmit array was based on the requirements of black-blood (BB) vessel wall imaging with DANTE preparation pulses [12]. Based on successful BB DANTE imaging parameters used at 3 Tesla, the target B_1+ field for 7 Tesla was determined. Sequence parameters that have worked at 3T include: Number of DANTE pulses: $N_p=150$; Pulse width: $P_w = 60\mu s$; Inter pulse spacing: $t_D=1.5ms$; DANTE Flip angle: $\alpha_D = 15^\circ$. Therefore, the Tx array was designed to ensure that any spins flowing towards the imaging plane would be subjected to at least 150 DANTE preparation pulses at the above flip angle. The extent of the region proximal to the imaging plane, with respect to the blood flow direction, was estimated using Equation (1).

$$L_{B1} = v_{blood} \cdot N_p t_D = 40cm \cdot s^{-1} \cdot 150 \cdot 1.5ms \simeq 100mm \quad (1)$$

In this equation, L_{B1} is the distance proximal to the carotid bifurcation where spin excitation is required; v_{blood} is the blood velocity in this region, N_p is the number of DANTE pulses; and t_D is the DANTE inter-pulse interval. The blood velocity (v_{blood}) in the aorta was assumed to be 40 cm/s, the maximum documented in literature [34]. The average velocity in the carotid arteries, however, is 35 cm/s and hence the above value was used for safety margin. Here, For the sequence specifications prescribed above the required L_{B1} is therefore about 100mm.

The DANTE BB effect has been shown to be effective for pulse flip angles of between 3° to 15° . For the most demanding scenario of 15° the required B_1 field strength can be estimated using Equation (2):

$$\alpha = \gamma B_1 \tau \Rightarrow B_1 = \frac{\alpha}{\gamma \tau} = \frac{15^\circ / 360^\circ}{42.6MHzT^{-1} \cdot 60\mu s} \simeq 16\mu T \quad (2)$$

where γ is the gyromagnetic ratio of 1H ; τ is the pulse width and α the DANTE flip angle.

This requirement was, however, reduced by a factor of two by doubling the duration of the DANTE RF pulses. Thus, $8\mu T$ minimum B_1+ for at least 100mm distance proximal to the bifurcation for the two common carotid arteries was the target specification. This profile was designed to be realised with multiple surface Tx coils.

For imaging of the carotid bifurcation itself, and DANTE preparation proximal to this region, two imaging transmit surface coils were designed to be located at each side of the neck, at the level of the bifurcation. Two more surface coils were placed on the lower part of the neck, approximately above the collar bone region, to ensure adequate B_1 spatial extent (Fig. 1). The two side coils were designed as two parallelogram-shaped loops on each side of the imaging plane with dimensions of 60mm to 90mm. The angle of the two parallelogram loops was 18° to accommodate the local anatomical restrictions imposed by the shoulders. Finally, the two top loops were circular and 80mm in diameter (Fig. 1).

The Tx coil configuration was designed around an anatomically realistic model from the virtual population provided as part of the Semcad-X EM simulation software (Speag, Zurich) [34,35]. In the simulations, an RF shim of 0° , 0° , 180° , 90° for channel 1, 2, 3 and 4, respectively, was used, and 1000W per channel, which was the maximum power per channel. The target value of $8\mu T$ was achieved in the simulations and therefore the construction was commenced (Fig. 1, left panel).

[Figure 1 approximately here]

RF safety limitations were not taken in account at the design stage but were incorporated after the B_1 target value was secured. Based on the coil RF profile and the energy deposition rate of the sequence in use, an appropriate power over time (10 sec and 6 min) limitation was set. The RF

safety approach was based on calculating the worst case of either global and local SAR values, and in most cases global SAR was the limiting factor.

2.2 Receive Array Design

Two overlapped loops at each side of the neck at the height of the imaging plane comprised the 4-channel receive array (Fig. 2). The diameter of each loop was 50mm so that the sensitivity, as well as the SNR benefit from using multiple loops, was adequate for a typical carotid artery depth below the skin, roughly 20mm [36]. The receive array was designed to be placed as close as possible to the neck with the two side Tx loops positioned about 30mm further out. Multiple loops were also used to allow implementation of parallel imaging schemes and hence facilitate a reduced scan time.

[Figure 2 approximately here]

2.3 Coil Construction

Decoupling between the transmit and receive arrays, as well as decoupling between the loops within each array, was performed by: (i) geometric decoupling by overlapping the loops where possible; (ii) active detuning of the Tx or Rx array by PIN diodes; (iii) passive detuning for the Rx array and (iv) pre-amplifier decoupling, a method that further isolates the coupling between receive loops in an array [38].

Geometric decoupling was employed for both the Rx and Tx arrays. Each two neighbouring loops were tuned to the Larmor frequency and then brought towards each other until the characteristic resonance splitting completely disappeared. Active detuning was deployed at each of the 8 loops with the use of PIN diodes (Temex DH 80106) which can withstand 500 W CW power. Biasing was delivered via RF chokes and DC isolation capacitors for the least possible interference. Passive decoupling was done via a pair of crossed diodes (Microsemi UM9989) with a small inductance on each of the Rx loops for more isolation during transmission. Circuit diagrams and implementation are shown in Figures 3 and 4.

Preamplifier decoupling was employed for each of the receive loops. The Low Noise Amplifier (LNA) used here had an S_{11} of $0.95\angle 48^\circ$ (vendor-provided preamps retuned at 297.2MHz) and its impedance was brought to an open circuit, in series with the loop, with the use of an appropriate coaxial cable length and capacitance (Fig. 3, 4) [39,40]. The performance of the isolation was assessed by an S_{12} measurement with a double probe and the peak difference was found to be 25dB or more in each case.

[Figure 3 approximately here]

[Figure 4 approximately here]

For avoidance of preamp oscillations, coupling between the two preamp ports was suppressed by the use of shield traps in the coax cables at both ends. DC cables can also carry RF and thus were modified in order to eliminate interactions. RF chokes in series with inductances of $2.2\mu\text{H}$ (Coilcraft, Cary, IL) were periodically added along the cable, also blocking RF from damaging the DC source. Large RF/DC isolation capacitors of 1nF (American Technical Ceramics, NY, USA) were inserted between the detuning circuits and the main loop. The same capacitors were inserted between the DC ground and DC bias cables, which were twisted to reduce the effective loop cross section (Fig. 5).

[Figure 5 approximately here]

A plastic casing was designed in SolidWorks (Dassault Systems, France) in ten pieces and was 3D printed with ABS material. The top and bottom of the coil base was CNC milled from 5mm Perspex sheets for overall robustness. The whole assembly was put together with M4 nylon screws. The top compartment hosting the two top transmit loops is removable so that the subject can lie between two sides of the coil. This compartment is connected via coaxial cabling and BNC connectors to the base for feeding the RF and the DC logic signal for the PIN diodes. One vendor-supplied 7T plug was used (ODU, Germany) for the 4 Rx channels, 8 PIN diode biasing (Tx and Rx) as well as coil coding (2 resistor code, 3 pins). A multi-channel Siemens Step 2 parallel transmit (pTx) plug was used (ODU, Germany) for the transmission path. Patient comfort was also taken in account during the design stage. The final coil is shown in Figure 6.

[Figure 6 approximately here]

2.4 Coil Bench Tests

The coil was evaluated on the bench using a Hewlett Packard 8712-C network analyser, loaded with a cylindrical water phantom of 120 mm diameter. The coil was treated as an 8-port system, one port for each channel of the Tx and Rx arrays, and the scattering parameters were measured for the two modes of operation (Tx and Rx). The isolation measurements were performed before the preamp connection in the Rx loops and therefore, the preamp decoupling was not taken in account. The Q-factor was measured for the Rx loop based on the -3dB points of the S_{11} mode. The measurement was done while the load was placed at distances of 5, 10, 15, 20 and 30 mm, while the most representative human loading distance was 20mm.

The numbering of the Tx array was Tx1 and Tx2 for the side loops, Tx3 and Tx4 for the top overlapped loops; Tx1 and Tx3 were on the left side of the neck while Tx2 and Tx4 on the right side. Similarly, Rx1 and Rx2 are the two left receive loops and Rx3 and Rx4 the right side loops.

Although not strictly a bench measurement, g-factor measurements are included here since they do not involve assessment of the coil with regard to blood suppression. For this calculation, the root sum of square SNR image was used to calculate the noise amplification for acceleration values of R=2 and R=2x2. A FOV of 120mm was selected to accommodate the diameter of the cylindrical phantom and restrict additional aliasing outside the selected phantom [41,42].

2.5 Flow Phantom Experiment

An experiment with a flow phantom was run for evaluation of the flowing spin suppression capability. The phantom comprised a 4mm inner diameter flexible plastic tube pumping water via a peristaltic pump. For a more realistic coil loading, a 2 kg meat phantom (bovine) was used (Fig. 7). The tested fluid velocities were 10, 20, 30 and 40 cm/s in order to include the expected statistical maxima and minima of the blood velocities in the carotid arteries. The coil was set to operate within safety power limits by adjusting the TR as required. The experiment was repeated for flow in both directions in order to assess the suppression on both sides.

Flowing spin suppression was evaluated by calculating the reduction ratio of the averaged signal in the lumen between the two flow directions (Fig. 7). For A being the signal from the unsuppressed flow direction and B the signal from the suppressed flow direction, the reduction was calculated as $(A - B)/A \times 100\%$.

[Figure 7 approximately here]

2.6 Human Scans

MR images from five healthy volunteers, under institutional ethical approval, were obtained for coil evaluation. In order to stay within the SAR guidelines, the Tx power had to be restricted to take into account simulations with human phantoms. Thus, a conservative RF supervision scheme was employed based on the worst-case scenario calculation. The reliability of the software was validated with experimental temperature measurements.

For each of the subjects a series of GRE scout (localiser) images were taken, followed by a C-spine anatomical image. A PD-weighted turbo-FLASH MPR (Multi-Planar Reformat) sequence was used with 220mm FOV; Base Resolution of 256 with 100% PE; slice thickness = 1 mm; TE/TR=3.44/1040 ms; BW = 179 Hz/pixel and R=2 parallel imaging. The total scan time was 4min 30sec.

The read-out used for BB DANTE prepared imaging was a 2D or 3D fast low angle shot (FLASH) sequence. The parameters of the DANTE module were set at $N_p=150$ pulses, $\tau=200\mu\text{s}$, $t_D=1.5\text{ms}$, $\alpha_D=15^\circ$ and 18mT/m gradient in every direction. The FLASH readout was implemented with centric k-space acquisition scheme for each of the segments. The time duration of each segments was approximately 80-120 ms. Parameters for FLASH were $\alpha=15^\circ$, FOV=150mm, base resolution=256, PE 100%, GRAPPA R=2, slice 2D: 2.5mm / 3D: 0.6mm, TE=3.68ms, TR 2D: 10s / TR 3D: 5s. As mentioned above, the pulse width was increased from $60\mu\text{s}$ to $200\mu\text{s}$ to ease the B_1 and therefore power requirements. Before each subject was scanned, an RF shim was performed on the imaging plane with the use of the installed algorithm in the Siemens 7T Step 2 scanner (Erlangen, Germany).

For a 15° flip angle in the carotid arteries, the required B_1 was found to be approximately $5\mu\text{T}$. The input voltage was calculated and set manually at $55 V_{\text{rms}}$ to achieve a mean of $5\mu\text{T}$ along the artery and a second option of $87 V_{\text{rms}}$ input was used to achieve $5\mu\text{T}$ in most of the artery ($5\mu\text{T}$ being the 'mean'-'std'). The choice of power input levels had implications for the rate of deposited power and thus the scan duration. After calculation of the voltages and the RF shim on the slice, the TR was modified in order to keep below SAR limits.

The restrictive nature of the SAR supervision scheme resulted in a 3D sequence of 12 min for R=2 which is an unusually long duration in a clinical setting. For the 2D protocol 12 slices were selected and the higher voltage was applied. The total duration of the 2D sequence was 4 minutes. Another 4 minutes of idle scan time were required between the two protocols for the SAR calculation to reset (effectively allowing subject 'cooling') and to avoid any scan interruptions from the RF monitoring system. Due to the long single sequence duration and discomfort from the 7 T environment (bore diameter, B_0 physiological effects) motion artifacts were expected.

The SNR was measured as the mean signal of the selected area divided by the standard deviation of a background ROI, free of artifacts where possible. The CNR was also calculated for the lumen vs wall in the images where blood suppression was applied. In particular the calculation was $(S_w - S_l) / \sigma_0$ where S_w is the vessel wall mean signal, S_l the lumen mean signal and the σ_0 the noise standard deviation.

Due to the use of SENSE with R=2, the simple method used to calculate noise is not strictly correct, as it fails to account for g-factor noise. This has to be taken in account when comparing the values with the 3T scans with no parallel imaging.

3 RESULTS

3.1 Coil Bench Tests

The scattering parameter results in dB of isolation (or reflection) are given in Tables 1 and 2. The reflection coefficient, and therefore the match on the load, is given for the corresponding mode and is marked with bold text in the tables. For the Tx mode, the match was below 15dB in each

case and therefore adequate for power efficiency. The two upper coils had a relatively poor isolation of about 10dB while the Tx1/Tx3 and Tx2/Tx4 coils, which are the loops on the same side, had isolations of 13dB and 18dB respectively. There was also fair (~17dB) to very good (~22dB) isolation between the Rx loops and the Tx coil on the same side. However, passive and preamp decoupling could not be measured and therefore the isolation would be higher during operation in the scanner.

The Rx match was below -22dB for each array element which was deemed excellent. The isolation values between Rx1/Rx2 and Rx3/Rx4 were about 13dB for both but ignoring any preamp decoupling. The isolation with the Tx array was more than 22dB, and therefore satisfactory.

[Table 1 approximately here]

[Table 2 approximately here]

Table 3 summarises the measured Q-factors for the Rx loop for different distances between the phantom and the coils. At the most representative loading distance of 20mm the Q was found to be 79 with the Q_u/Q_L being 3.4, showing adequate loading conditions.

[Table 3 approximately here]

The g-factor map measurements for the Rx coil are shown in Figure 8. With 4 channels placed in two opposite pairs it would be unreasonable to expect good performance for $R > 2$ since the opposite coil sensitivities correlate very little. Therefore, an $R=2$ factor was used for reduction of scan time.

[Figure 8 approximately here]

3.2 Flow Phantom Experiment

The quantified flowing spin suppression results are summarised in Table 5 where the average suppression over 4 measurements was 78.8% for 10cm/s, 80.3% for 20cm/s, 77.3% for 30cm/s and 73.9% for 40cm/s, showing a robust performance over flow velocity. The total average suppression was 77.6%, which was deemed adequate to proceed with human scans. Images and visual evidence of the suppression is provided for each of the 4 different fluid velocities and directions (Fig. 9).

[Table 4 approximately here] [Figure 9 approximately here]

3.3 Human Scans

The PD-weighted turbo-FLASH MPR sagittal images of the cervical spine, taken from one subject, gave an average SNR of 92 ± 5 in the CSF and 80 ± 4 in the vertebrae (Fig. 10). In a selection of the most artifact-free images from the 2D protocol, the SNR for the tissue close to the skin was found to be 56 ± 6 ; the SNR close to the vessel wall was 24 ± 4 ; the SNR of the lumen was 10.9; and the CNR between lumen and adjacent tissue was 17.3 ± 2 (Fig. 11). Similarly, for the 3D protocol, the tissue SNR was found to be 55 ± 8 ; the vessel wall SNR was 33 ± 6 ; the lumen SNR was 11.5; and the CNR between lumen and surrounding tissue was 25 ± 4 (Fig.12).

Finally, DANTE black blood images were taken for comparison at 3T with DANTE parameters as mentioned in Methods section and a 3D FLASH read-out for fair comparison. The RF coil used at 3T was a 2-channel Rx in combination with the standard transmit body coil. The tissue SNR

was measured at 54 ± 8 ; the vessel wall SNR at 20 ± 5 ; the lumen SNR was 10 ± 1 ; and the CNR was 13 ± 2 .

[Figure 10 approximately here]

4 DISCUSSION

An MRI coil was designed to satisfy the RF field requirements of black blood imaging with DANTE preparation pulses. The coil comprised a 4-channel Tx array and a 4-channel Rx array. The receiver array allowed use of parallel imaging for scan reduction time. To ensure RF safety, a worst-case scenario method was adopted and realised via EM simulations. After phantom testing, healthy volunteers were scanned for black blood and anatomical imaging and the results indicated that the constructed coil achieved the required B_1^+ . However, motion artifacts due to long scan time degraded the quality of the images.

With regards to flowing spin suppression, it could be argued that the suppression is more dependent on the coverage of the transmit field than on the Larmor frequency and the intrinsic magnetisation. A clear advantage of the 3 T field strength is that the whole-body transmit coil coverage allows for long DANTE pulse trains, ideally of low power, and hence affects the static tissue less. At clinical fields, without parallel Tx arrays, it is possible to run SE read-out sequences, since SAR can be effectively monitored. The restrictive power regime based on the worst case safety limits made the use of SE or TSE readouts almost impossible. However, on-the-fly accurate safety supervision regimes could soon become available and the power efficiency of Tx arrays at 7 T should increase. Having comparable black blood imaging quality for the same parameters is nevertheless a step forward at 7 T, showing the feasibility of the study and the potential of the application at UHF strengths.

The high-resolution readout used in combination with the DANTE module gave an average lumen SNR of 11 for both 2D and 3D reconstruction protocols; the average tissue SNR was 55; the average vessel wall SNR was 29; and the average CNR was 21. Previous data from 3T DANTE BB imaging had a lumen SNR of 10; static tissue SNR of 54; vessel wall SNR of 20; and an average CNR of 13. Even though a small SNR/CNR improvement is observed at 7 T, it is not proportional to the field strength as expected. Aside from the fact that the comparison was done between different coils, the 7 T images suffered from aliasing from parallel imaging and motion artifacts due to long sequence time, dictated by the power restrictions for RF safety.

A further improvement to the approach would be a more effective RF shimming strategy for the whole DANTE volume rather than just the imaging plane (i.e. a single slice, as was acquired here). However, the lack of receive elements over a wider volume would also need to be addressed. Finally, B_0 shimming, either static or dynamic, could be employed to reduce the susceptibility effects and T_2^* decay.

5 CONCLUSION

The results from phantom and subject imaging show satisfactory blood suppression for the whole cross sectional velocity profile. For the phantom measurements, the signal reduction of the circulating fluid with DANTE pulses was at least 74% for 40 cm/s velocity, typical for the diastolic phase in the carotid arteries. In human subject imaging, proton density c-spine imaging gave a relatively high SNR of 90. However, UHF could benefit further from emerging accurate RF supervision schemes.

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