Optimization of carotid coils

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Introduction: The vulnerable plaque has a thin fibrous membrane, of less than 65µm, covering a large lipid, possibly necrotic core. A necrotic core of the plaque can be on the order of 1mm^2. A rupture of this thin fibrous cap exposes the thrombogenic lipid core of the atheroma to flowing blood, leaving the patient with a high risk of thrombosis. For accurate determination of vulnerable atheromas, extremely high spatial resolution will be required to measure the necrotic core, thrombolytic element and possibly the thickness of the fibrous membrane. . Detecting plaque inflammation due to macrophage infiltration may also provide important information in the diagnosis and treatment of vulnerable plaques. This type of high resolution, MR imaging, will require sufficient SNR. To provide this SNR, a plethora of RF coils geometries have been evaluated for imaging at the level of the carotid bifurcation.

Method: In order to optimize the carotid coils two-, four-, six, and eight arrays with different size and geometries where constructed and relative SNR compared at 3.5 cm depth (approximate position of carotid artery bifurcation from the surface of neck) using a cylindrical SNR phantom. Due to the depth penetration and decoupling issues small coils were unsuited for this purpose even though they provide high SNR near the coil. Fig A shows the relative size and position of the coil with respect to the human neck. Ten different coils were compared: (1) 4 element coil (10cmX10cm), (2) single rectangular coil (10cmX10cm), (3)2 channel rectangular coil(10cmX10cm), (4) 4 channel rectangular coil(12cmX10 cm), (5) 3" circular loop, (6) 4 element coil (15cmX15 cm), (7) 3"circular loop with a butterfly on top(8.2cmX8.6cm), (8)two 3" circular loops, (9) 2 channel birdcage coil (15.24cmX7.62 cm), and(10)Three 3" loops, on each side. Overlapping and inductive decoupling methods were used to decouple the adjacent and non-adjacent elements respectively. All coils were tuned and matched to 50 Ω at 63.86 MHz, while loaded with the phantom and were attached via baluns to half wavelength cables to low input resistance preamps, which further minimize the interaction between any surface coil and other surface coils not immediately adjacent thereto. A passive decoupling circuit was attached to each coil to decouple from body coil during transmission. We used a fast spin echo sequence for the comparison (256X256, TE=14ms, TR=300ms, FOV=20).

Figure A – 6 Channel Carotid Coil

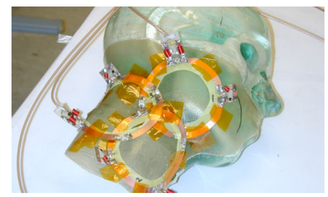
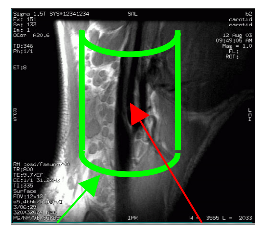


Figure B – Relative Coil Placement



Point of interest-3.5cm away from surface

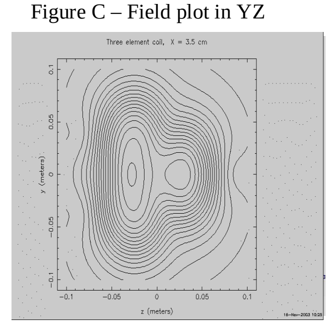
Comparative analysis was completed using human volunteers under appropriate IRB conditions for two black blood based MR techniques. Both measurement utilized a double IR FSE techniques with a 256 x256 matrix, FOV 14cm, 1 NEX, 2mm slice thickness, ETL 10 and BW of ± 64 kHz. The TR was set by using 2-RR interval, which was approximately 1600ms for volunteer imaging. TE on the proton density weighted images was 6ms and 58.8ms on the T2 weighted images.

Result and Discussion: Table 1 shows a comparison of SNR of all coils at a depth of 3.5 cm, and it is obvious that the 6-channel coil performs better than all other coils. The number of coils not only produced additional superficially, but also were large enough to produce the needed level of penetration. Additionally, the 3coils on each side of the head enabled a large enough area cranial to caudal to ensure the coil did not need to be moved for patient variation.

Table 1: SNR comparison at 3.5 cm depth between different coil geometries

Coil	1	2	3	4	5	6	7	8	9	10
SNR	422	420	542	543	589	587	676	721	587	730

Figure C – Field plot in YZ



The field plot in figure C shows the 6-Channel carotid coil to have excellent homogeneity, which will allow fat saturation techniques better performance. Table 2 shows the comparison between the 6 channel coil (10) and the 3"loop (5) in proton density and T2 weighted images. The 6 channel coil (see Table 2) shows a significant improvement of SNR and CNR over the 3" loop. Increasing the number of coils over this small region (by eight or more than that) may increase the SNR superficially, with a penalty of depth penetration and decoupling. Images such as the one displayed in Figure D (proton density image) show good penetration of signal at the level of the bifurcation. Although, some coil 'burn-in' was noted, it did not interfere with the conspicuity of the carotid wall or lumen. Dramatic improvements in CNR were also seen at the level of the wall.

Figure D- Proton density image

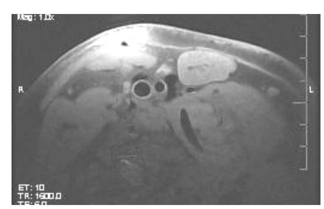


Table 2: 6-channel coil (10) compared to 3' loop (5) for Proton Density and T2 images

	6 Channel PD	3 inch Loop PD	6 Channel T2	3 inch Loop T2
SNR (wall/ noise)	28.45	16.39	19.22	11.86
SNR (fat/noise)	13.64	4.43	15.07	5.9
CNR (wall-blood)/noise	24.19	14.09	15.07	9.07
CNR (wall-fat/noise	14.81	11.96	7.78	5.96
CNR (fat-blood)/noise	9.38	2.13	7.29	3.11

Conclusion: Initial experimental results the 6- channel coil offers the advantage of additional SNR/CNR, while maintaining depth of penetration necessary for carotid imaging. Also, parallel imaging would allow nearly a factor of three decreases in time using the phase difference between the coils. This will be important for the longer imaging times require for high-resolution imaging.

Reference: .1. *Bruce A Wasserman, et al.* Radiology 2002; 223:566-573