Mixing loops and electric dipole antennas for increased sensitivity at 7 Tesla

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Introduction: By using a current mode expansion employing dyadic Green's functions (DGF) [1,2] it is possible to calculate the Ultimate Intrinsic SNR (UISNR) for different points within a uniform cylindrical or spherical object. The current is defined to flow on a surface a given distance above the surface of the object. For high frequencies and large objects, curl-free current modes (corresponding to electric dipoles) make a significant, even dominant, contribution to the UISNR for the center of the object [3,4]. Using pure electric dipole elements to excite and/or detect an MR signal is something that has only recently begun to be explored. We investigate here through simulation and the construction of coil prototypes the use of mixed loop and electric dipole elements to provide improved SNR for body imaging at 7 Tesla.

Methods: A large phantom was constructed enclosing a cylindrical liquid volume 29.5 cm in diameter and 117 cm in length. This was filled with deionized water with 1.24 g/liter of $NiSO_4x6H_20$ and 2.62 g/liter NaCl, resulting in $\varepsilon_r = 80.8$ and $\sigma = 0.604$ s/meter at 297.2MHz, measured with a dielectric probe and network analyzer (Agilent, Santa Clara CA). These dimensions and properties were used to calculate UISNR at 297.2 MHz for the center of the phantom with a current surface 1 cm above the phantom [1,2]. This was done for three cases: only divergence-free current modes (loop-like), only curl-free current modes (electric dipole-like) and allowing all current types.

Several practical coil configurations were simulated using the finite difference time domain method (Microwave Studio, CST, Framingham MA). Using the same dimensions and properties as the constructed phantom, coil elements were modeled on a cylindrical surface 1 cm above the phantom. Two loop arrays were simulated with rectangular elements 14.6 x 15 cm in dimension, overlapped to null mutual inductance with neighboring coils, in a single row of 8 (Figure 1a). A single row of 8 electric dipoles was modeled, with each element self resonant with a conductor length of 35.6 cm (Figure 1b). Fig. 1c shows a 16 element array consisting of a single row of 8 overlapped loops with a 35.6 cm long electric dipole centered over each loop. Finally a single row of 16 overlapped 7.3 x 15 cm loop elements was simulated (Figure 1d). All coil elements were tuned and matched in the simulation and driven with 50 Ohm ports.

A coil array was constructed to closely match several of the simulated coils. Constructed on a 31.5 cm outer diameter acrylic tube, it consisted of 8 overlapped loops and 8 electric dipoles. The loops were 14 x 15 cm constructed of tinned bus wire incorporating 11 distributed capacitors and matched to coax with $\lambda/4$ lattice baluns. The dipoles were constructed from FR4 circuit board with 7 mm wide traces with lengths adjusted between 32 cm and 36 cm to fine-tune them according to their proximity to the phantom. All elements were matched to coaxes with $\lambda/4$ lattice baluns. The coil was connected to the scanner using in-house constructed T/R coil interface boxes. For dipole-only and combined dipole-loop experiments the dipoles were used to transmit. For dipole-only or loop-only experiments the unused coil elements were defeated by removal of capacitors or by cutting through the dipole conductors. In each experiment equal power was supplied to each channel and phases adjusted to provide circularly polarized excitation at the center of the phantom

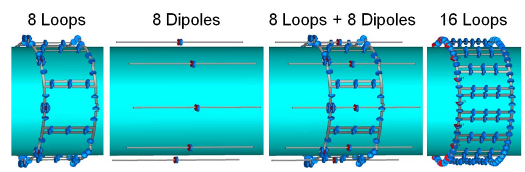

Fig. 1: Coil structures simulated with FDTD method

All data were acquired on a 7 Tesla whole body scanner (Siemens Medical Solutions, Erlangen, Germany) with an 8 channel parallel transmit system. SNR maps for the optimal combination [5] were generated from GRE acquisitions with and without RF excitation (TR/TE/Flip/BW = 1000/3.39/20/300, FoV=400mm, Matrix=128, Slice = 5mm) after calibrating the excitation flip angle at the center of the phantom. B_1^+ maps with matched FoV and matrix were generated using the AFI technique [6], acquiring maps with transmit phases corresponding to the "uniform" birdcage mode and then with the first gradient mode, as well as corresponding low flip angle GRE magnitude images. Normalized SNR maps were obtained by dividing the SNR maps by the sine of the measured flip angle at each pixel.

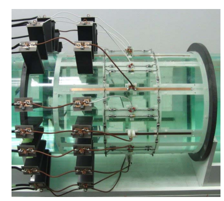

Fig 2: Constructed coil

Results: Derivation of the UISNR for the center of the phantom shows that curl-free (dipole-like) currents provide 21% higher SNR than loop-like currents, and that allowing for both types of currents provides 54% higher SNR than loop-like currents alone. CST simulations achieved S_{11} match of better then -18 dB on all ports and maximum S_{12} coupling of -13.7 dB. Simulation of 8 dipoles alone provided a 3.1% boost compared to 8 loops, but the combination of the two provided an SNR boost of 24% (Figure 3). This additional SNR cannot be attributed simply to the increased number of elements in the combined array, since the simulation with 16 loops only provides a 1.2% SNR increase compared to 8 loops. For the constructed coil the overall pattern of SNR is very close to the simulated results. Experimental results with 8 dipoles alone provide an 8.3% boost compared to 8 loops, higher than the 3.1% found in simulation. Using combined loops and dipoles provided a 22% boost, very close to the 24% boost found in simulation.

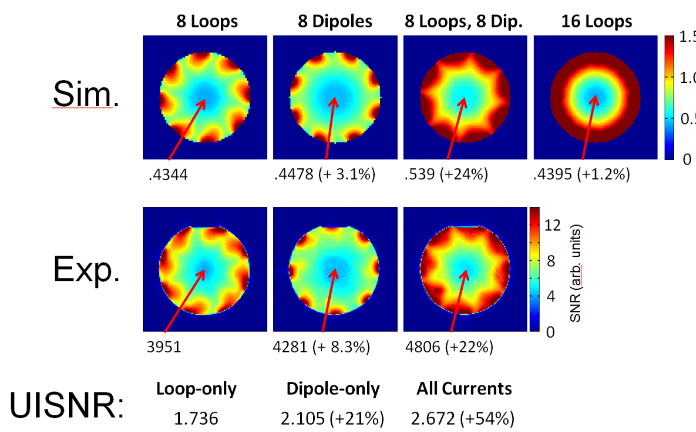

Figure 3: Optimum SNR maps, normalized for B_1^+ distribution, and values of Ultimate Intrinsic SNR for this geometry (arbitrary units).

Discussion: Analysis of UISNR suggests that SNR might be increased by up to 21% by the use of dipoles instead of loops, but we see a more modest increase in the simulated and constructed coils. Close inspection of the ideal current patterns associated with the UISNR [3] in each case suggests that this may be attributed to the fact that for this phantom size and frequency the curl-free currents do not form simple linear dipole structures but have considerable phase evolution along Z, and hence the constructed coil does not overlap strongly with the actual ideal current pattern. SNR at the center of the phantom approaches divergence-free UISNR rapidly as the number of loop elements is increased [3], as evidenced by the modest SNR increase for 16 compared to 8 loops in simulation. Thus the 22% SNR increase provided by the combination of loops and dipoles is significant, and coil arrays incorporating this principle should provide increased SNR for body imaging at 7T. Further gains might be obtained by developing coil structures which more closely mimic the ideal current patterns.

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