Efficiency of a 3T whole body 16 channel TEM transmit array

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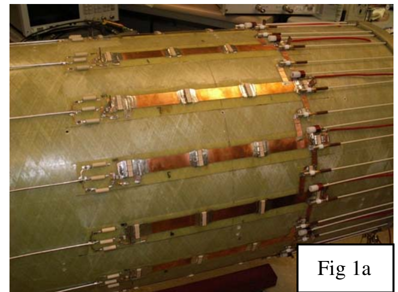

Fig 1a

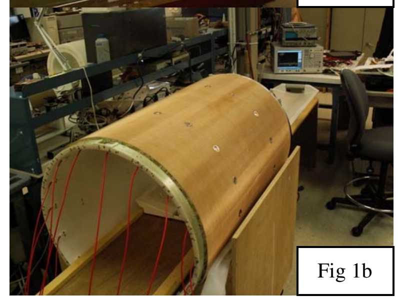

Fig 1b

INTRODUCTION

Body imaging at 127 MHz (3T protons) has shown significant inhomogeneities in the RF transmit field due to wavelength effects and permittivity of tissue. Proposed techniques to counter these effects like B_1 shimming and parallel transmit all require a whole body transmit array coil. This transmit array can be either an array of loops (1), an array of TEM elements (2), or a degenerate birdcage (3). Since each element in a whole body transmit array excites only a small portion of the volume, and is usually much closer to the RF shield than to the patient, coil losses are dominant, and significant RF power is needed per element to create B field in the center of the coil volume. It is the purpose of this investigation to compare the power efficiency of different whole body transmit array designs.

METHODS

A 16 channel 3T whole body TEM array was built based on a scale model study (1), FEM analysis (HFSS, Ansoft), and circuit analysis (ADS, HP). Each TEM element consisted of 1 inch wide, 42 cm long copper strips with 4 junctions of 36 pF (fig 1) to create a 3dB 40 cm FOV in the Z direction. Each junction consisted of 3 E style capacitors in parallel for better current handling. All strips were mounted on a fiberglass tube with an OD of 610 mm. A honeycomb polycarbonate structure over the top of the strips supported a copper mesh RF shield with a diameter of 650 mm. The mesh contained 325 lines per inch and has high impedance for gradient eddy currents. Elements were decoupled from direct neighbors by means of a capacitor between the rungs. All functions: tuning, matching, and isolation were implemented via tunable elements with remote access, such that the RF shield did not require removal during the tuning process. Pin diode decoupling networks needed for surface coil imaging, were implemented opposite from the input networks to keep RF and DC cables separated. Cable shields were connected to the RF shield to reduce common mode interference. The coil was integrated into a 3T MRI system with 8 independent transmit channels. Each amplifier was connected to the coil either via a butler matrix network or was connected to 2 strips via a splitter. These 2 strips were either adjacent (22.5 degree splitter) or opposite (180 degree splitter). The coil was replacing a whole body 8 channel loop array described earlier (1) and was compared to that coil with respect to temperature rise, and power efficiency. Temperature was measured on the ID of the patient bore using a fiberoptic system (Neoptix) at a power level of 400Watts 5% duty cycle net forward power into an empty coil element. The efficiency was measured using a flip angle mapping routine at a known amount of net forward power at the coil inputs.

RESULTS and DISCUSSION

A circuit simulation of the entire circuit had shown that isolation optimization between direct neighbors using a tunable capacitor worsened the isolation between next to nearest neighbors (fig 2), hence the optimization point was chosen at the point where those isolations were similar. Another ring with isolation capacitors can optimize both. The S parameter matrix was measured with the coil mounted in the MRI system, loaded with a 200 lbs volunteer. Reflection coefficients were all smaller than –20 dB, and isolation between any neighbors was better than –14 dB, mainly determined by direct and next neighbors. The Q of the coil elements empty were all around 260. When loaded with the 200 lbs volunteer the Q of most elements dropped by 10% except for the rungs adjacent to the shoulders of the volunteer, where the Q dropped by 35%. The FEM analysis showed that 40 A per rung in a sinusoidal distribution of current phase (emulated birdcage mode) would generate 17.5 uT of B^+_1. The efficiency was measured in the 3T system using an elliptical phantom of dimensions 33 by 26 by 35 (z) cm filled with water doped with 5mMol $NiCl_2$ and 64 mMol of NaCl, but a silicon oil sphere of 26 cm diameter was also used. Corrected for reflected power 530 Watts was transmitted into a single rung of the array, and the flip angle map recorded (fig 3). From this data we calculated that that the efficiency is 0.231 uT /√W of B^+_1 in the center of the unloaded coil with all rungs receiving the same 1/16 th of that power properly phased to set up a homogeneous birdcage mode. This should be compared with 0.189 uT /√W for the 8 loop array(1), and 0.14 uT /√W for the 8 channel TEM reported in (2).

The bore temperature was measured over time using 5% duty cycle, and a power level of 400 W into a single rung. After thermal equilibrium (80 minutes) the rise in temperature was 16 C compared to 50 C for one element of the 8 loop array (1) operating at 2000W and 5% duty cycle. During the experiment all air cooling was shut off.

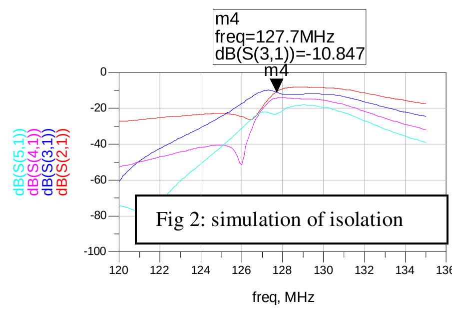

Fig 2: simulation of isolation

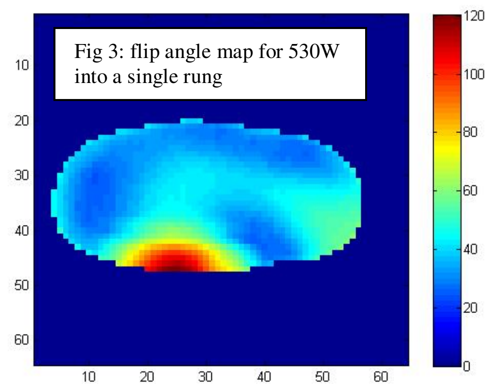

Fig 3: flip angle map for 530W into a single rung

CONCLUSIONS

The TEM design wins from the loop design because there is no similar cancellation of B field from neighboring conductors. The 16 element TEM wins from the 8 element TEM because most losses are coil losses and a 16 element coil requires half the current per rung to set up the same field. Even though twice as many rungs are needed, the power dissipation per rung is ¼, hence the 16 element coil will perform better if coil losses are dominant. In general transmit arrays with this shield distance are inefficient, coil losses dominate resulting in a rise in coil temperature. This temperature can be reduced by increasing the number of parallel capacitors per junction. Better arrays can be built by increasing the coil to shield distance, and by going to a TEM style array with as many elements as possible. If the number of transmit channels is smaller than the number of array elements, several elements can be combined via a splitter and phase shifter and still result in an efficiency increase over the smaller array element count.

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