An Elevated Endring Birdcage Coil for Improved Performance at 3 Tesla

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INTRODUCTION

Signal-to-noise (SNR) limitations of low field MR imaging has driven development of clinical high field systems (3.0T or greater). However, as the strength of the B0 field increases, the Larmor frequency of the B1 field increases proportionally. At such frequencies (128MHz and greater), the interaction between the B1 field and the patient cannot be neglected. This interaction is caused by the effective wavelength of the B and E fields, at higher frequencies, being comparable to or smaller than the dimension of the human body. Such a strong interaction substantially degrades homogeneity of the B1 field, negatively affecting image quality and causing an increased tendency for image shading. Limits on specific absorption rate (SAR), including new standards allowing 4 W/kg, have escalated concerns of patient safety, thus require efficient body coil designs utilizing E field reduction techniques.

METHOD

Simulations were performed for several body coil configurations, with the main objective being to minimize the E fields as relating to local SAR. At the same time, addressing inhomogeneity problems common to 3T. Special attention was given to reduce parasitic effects due to small values of tuning capacitors, center frequency shifts and isolation due to asymmetric patient loading. Coil prototypes were evaluated based on these performance criteria.

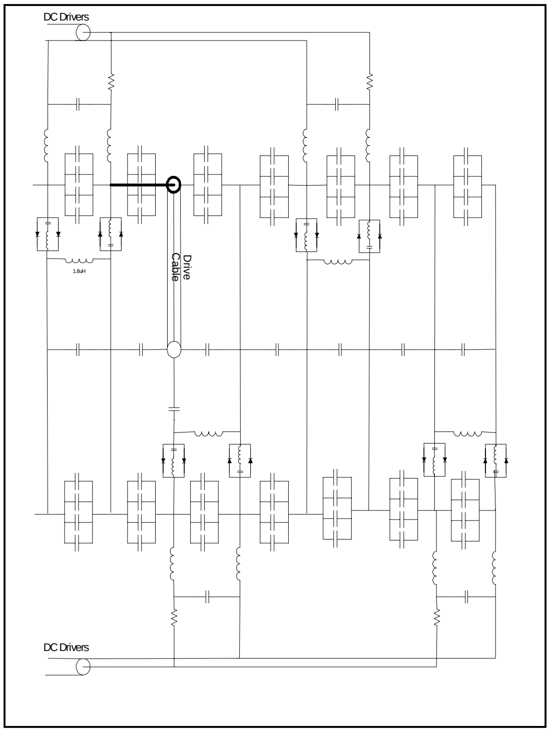

Figure 1: Partial schematic representation

RESULTS and DISCUSSION

A 60cm cylindrical high-pass birdcage coil was built utilizing 32 rungs with twice the width of the gaps. Widened endrings were elevated to a radius 1cm greater than the legs. A segmented RF fingerprint shield pattern, 65cm in diameter was used in conjunction with the coil. Current was distributed throughout the width of the endrings using multiple capacitors in parallel totaling 93pf (13.36Ω at 128MHz). Low impedance (<1 Ω at 128MHz) capacitors connect a third ring centered between the endrings at the central XY plane of the coil, forcing the symmetry about the coil. Peak currents in the rungs were 5 times lower than those in the raised endrings per Kirchhoff's current law. The coil was driven on the superior endring at two orthogonal ports. Coil Decoupling was achieved using series PIN diodes in each of the 32 rungs, as seen in figure 1. The junction capacitance of the PIN diodes were resonated using parallel inductors to achieve good isolation in each rung. Due to the addition of series resistance from the PIN diodes in each rung of the coil, unloaded Q values were only 175. Loaded Q values were still substantially lower at 35, providing an acceptable Q ratio of 5.

Less than a 400kHz shift in RF center frequency was measured for a large variety of patient loading, confirming low parasitics and E-field coupling to the patient. This is shown in figure 3, with the higher B1 efficiency compared with a smaller 16-rung birdcage design. Dielectric heating was measured directly at different locations around the inside surface of the RF coil and again compared to 16-rung designs. Locations nearest the endrings and capacitors observed the greatest heating; dramatic improvements were seen using this new design.

B1 homogeneity along the z-axis improves as a function of the endring's distance to the shield. As the endrings move closer toward the RF shield, the longitudinal B1 field is negated, and only the transverse B1 field remains. This benefits the homogeneity in the transverse plane, nearest the endrings, and reduces the slope of the B1 field away from isocenter in the coil, along the z-axis.

References

1. Watkins, R. et al. Proceedings of ISMRM, pg 1123, 2001.

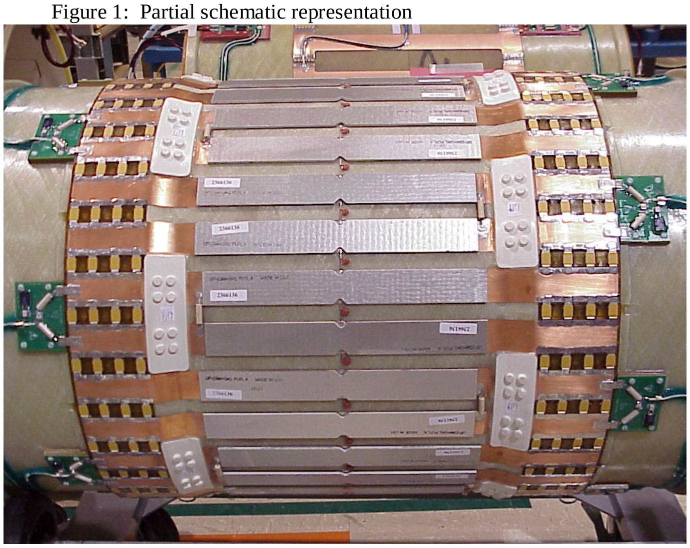

Figure 2: Picture of Elevated Endring RF Body Coil

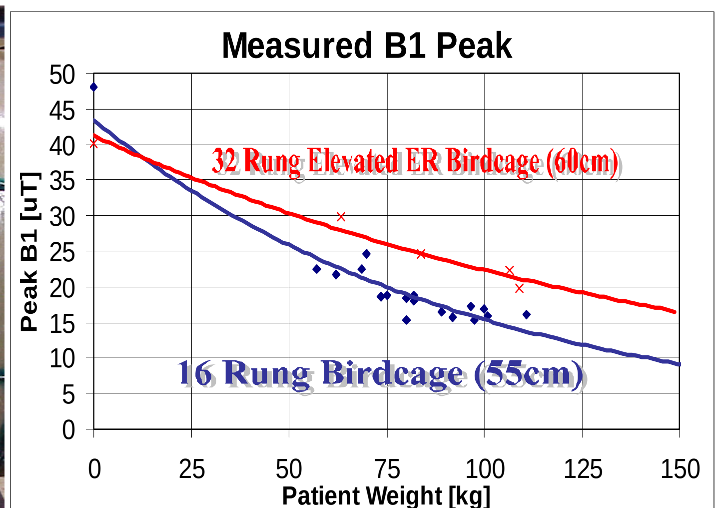

Figure 3: Maximum B1 using a 25kW transmitter for two body coils