A Novel Concept for Gradient Coil and Magnet Integration

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Introduction

As the main field magnet dominates any MRI scanner in size, weight and value, substantial progress in making MRI systems more patient friendly and less costly necessarily includes re-optimizing the magnet. In the following, a novel concept for a significantly closer integration of the magnet with the gradient coil (and the RF body coil [1]) is proposed.

Theory

Any cylindrical solenoid main field magnet consists of a set of superconducting coils with similar radii. Typically, the end coils carry the largest amount of wire. A reduction of the radius of these coils shows a triple benefit: The coils get closer to the imaging region, become more efficient and need less turns, the wire length per turn reduces, and the coils would also move longitudinally inward, resulting in a shorter magnet. The inner coils do not show this tendencies, so there is opportunity to reduce the radius of the outer coils at the expense of even increasing the radii of the inner coils. The result is a barrel-shaped inner magnet bore. Such a geometry nicely fits to the requirements of the flux return space inside the gradient coils, i.e. the optimum longitudinal distribution of the space between the gradient coil primary and its shield: The largest magnetic fluxes in the return space occur around z=0, and minimizing the field energy (inductance) necessitate large return flux cross-sections in the middle of the magnet. A similar argument applies to RF body resonators.

A further significant magnet size reduction - while still maintaining a mechanically separate gradient coil - can be achieved by shielding the magnet from the AC gradient fields not via the usual three gradient shield coil layers, but via the bore tube itself. Of course, the thickness of the conducting bore surface has to exceed the skin depth of the impinging AC fields.

Experimental results

A prototype integrated magnet and gradient system for a nominal patient bore of 60cm and a field of view of 30x30x30cm was designed and built. The magnet length was 1.04m instead of ~1.25m for a conventional design. The barrel-shaped 15mm aluminium bore tube had a diameter of 73cm at both ends and 93cm at the magnet center. The inner shape of the bulge was found by numeric optimization, matching the eddy current fields to the gradient coil magnetic fields inside the imaging region, and proved to be satisfactory for all three gradient axes simultaneously. The residual higher order eddy current fields not compensable by gradient current pre-emphasis were less than our standard requirements specs (<0.02%, <1µT) even with all manufacturing tolerances. Bore tube heating and associated thermal eddy current drifts were very low. The gradient coils had a G_{max} of 40mT/m and a slew rate in excess of 400 T/sm with standard 500A, 2kV gradient amplifiers, and needed ~38% pre-emphasis for the x and y axes, and 48% for the z axis.

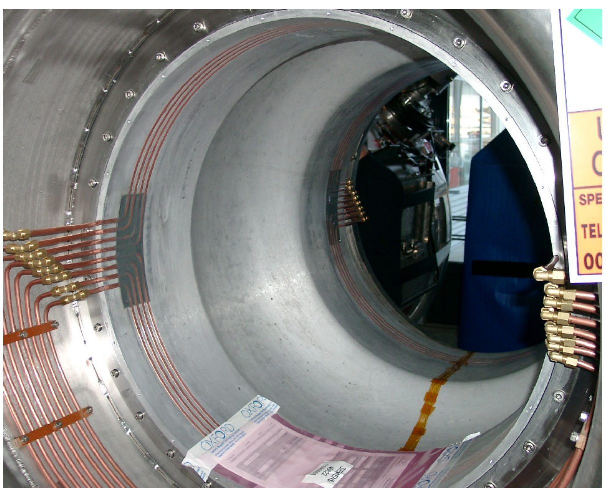

fig. 1: the bulged magnet bore tube

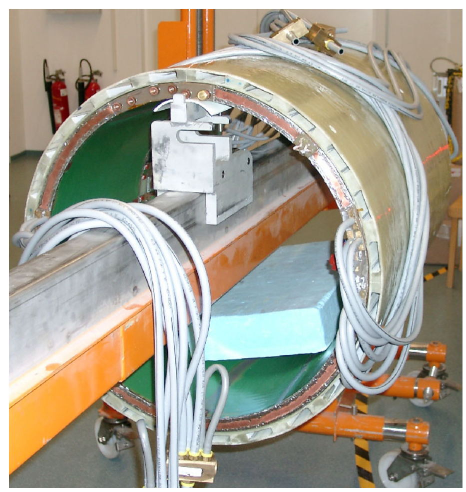

fig. 2: unshielded gradient coil with shim support

Conclusion

It is possible to significantly reduce the size of the main field magnet by close integration with the gradient coils (and the RF body resonator [1]). Additionally, practical experience suggests that gradient coils can be shielded passively, and the spatial distribution of eddy currents can be controlled sufficiently by shaping the shield current carrying surface, i.e. the magnet bore tube in this case.

References

[1] M. Vester et al. A novel concept for RF coil and gradient coil integration, submitted to ISMRM 2005