System for MRI guided Radiotherapy

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INTRODUCTION

The adverse side effects of gamma-ray radiotherapy can be reduced significantly by combining the treatment system with a real-time imaging system, used to precisely steer the radiation beam. MRI is a particularly attractive imaging modality for this, due to its excellent soft tissue contrast. A project is underway to combine a cylindrical, closed bore high-field MRI system with a 6 MeV Linac radiation source (fig. 1) [1]. A key feature of the concept is that the Linac is located on the outside of the MRI magnet. The gamma beam passes through the scanner to reach the patient inside the bore. This abstract describes some of the modifications to the MRI scanner needed to make it compatible with this application.

METHODS/DESIGN

The MRI system has to be modified in several respects to allow co-operation of a linear accelerator:

1. The magnet, gradient coil and all other system components in the central plane of the system has to be transparent to gamma rays. A number of relatively thin cylindrical walls, such as those of the outer vacuum container of the cryostat, are acceptable but superconducting coil bundles or thick gradient coil conductors would cause too much (inhomogeneous) absorption and scatter. The required width of the radiation window is approximately 150 mm at the windings of the magnet and 200 mm at the inner boundary of the gradient coil, preferably extending over (nearly) the full circumference of the system.
2. The stray field of the magnet has to be minimized along the path of the electron beam in the Linac, in particular at the gun section. This low-field zone can also be used to locate other parts of the Linac (e.g. a microwave circulator) that would not work in a strong field.

The magnet requirements can be fulfilled by not too drastic modifications of a conventional 6+2 section 1.5 T MRI magnet design. By shifting the shield coils towards the midplane and slightly modifying the number of turns of these coils, it is possible to create a ring-shaped field-free zone at a radius of 1.7-2.0 meters (Fig. 2). It is not easily possible to extend the low field region inward to the outer surface of the cryostat. The multi-leaf collimator of the Linac must be made from non-magnetic materials. Sensors and actuators should either be made field-resistant or moved radially outward. In order to retain the magnet's homogeneity on increasing the central gap width, a pair of small additional coils can be added, adjacent to the central sections. Alternatively, the inner radius of the central coils may be increased. The field pattern generated by the optimized split magnet design does not differ significantly from that of a standard magnet. A complete split of the cryostat, with vacuum-insulated walls bounding the central gap, would require a separation of the central magnet sections by at least 250 mm. Such a large separation would involve a major redesign of the magnet, leading to significantly increased conductor cost. The heat load on the cryostat by the gamma beam passing through is negligible, so the cryogenic performance and quench stability of the magnet will not be impaired by the combined operation.

The gradient coil has a central gap free of windings of 200 mm width. The main and shield saddle coil conductors of the transverse channels are interconnected by conductors running over the flanges bounding the central gap [2,3]. The coil also features some radial interconnections at the outer flanges of the coil (Fig. 3). The coil was designed by modeling the stream function on a surface mesh, minimizing the coil's magnetic stored energy while satisfying constraints on primary gradient field in the imaging volume and field contributions of induced eddy-currents [4]. With inner/outer diameters 700/850 mm the stored energy of the transverse channels is 7.8 Joule at a gradient amplitude of 10mT/m. This is 40% more than of an un-split coil with the same size and performance. Although there are no shielding conductors in the gap area, the field contribution from eddy currents induced in the magnet bore tube is of the same order of magnitude as seen in un-gapped actively shielded gradient coils. Both coil halves were manufactured using standard materials and manufacturing technology. The two halves are rigidly linked by means of a 5 mm thick GRP cylinder located on the outer periphery of the coil. This interconnection is thin enough that gamma ray absorption/scatter is acceptable but provides sufficient stiffness to the cylinder to withstand the Lorentz forces acting on it.

The RF coils and RF shields located inside the gradient coil do not contain any thick copper parts or lumped components in the region where the gamma beam passes through.

RESULTS

First tests of simultaneous operation of the modified MRI system and a Linac gamma source radiating through the magnet are expected to be carried out in the first half of 2009.

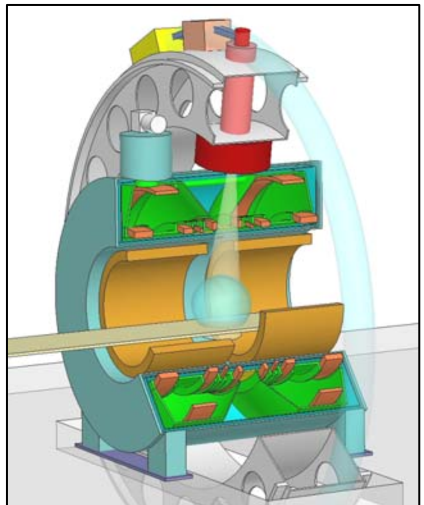

Fig. 1 combined MRI/Linac system

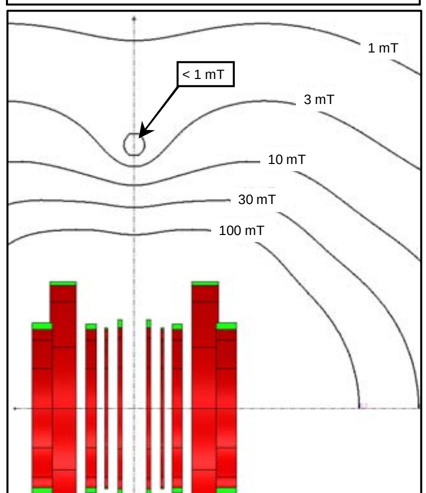

Fig. 2 low field region outside magnet

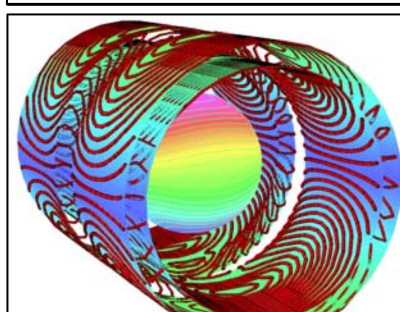

Fig. 3 split gradient coil

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