Design of a Whole-Body Radio Frequency Coil for image-guided radiotherapy treatment in a MRI-LINAC system

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INTRODUCTION: Radiotherapy is a primary cancer treatment modality. However, targeting a radiation beam on tumours in moving organs such as liver, stomach or lungs, without damaging healthy tissue can be challenging. For this reason, Magnetic Resonance Imaging (MRI) and Linear Accelerator (LINAC) are being combined as a MRI-LINAC, a new image-guided radiotherapy treatment capable to both image and accurately treat the patient simultaneously. In previous work [1-3], a 1T MRI magnet and gradient system with a large central gap (50 cm) was introduced, wherein the LINAC and the patient can be placed in an 'in-line' (Figure 1, left) or a perpendicular configuration (Figure 1, right**)**. Both configurations are considered, as they have different advantages and disadvantages [3]. A radiofrequency (RF) transmit volume coil was developed to fit into this specific system. The combination of MRI with radiotherapy introduced new constraints that had to be considered in the design of the coil. An unobstructed path for the LINAC beam to target the patient had to be combined with the requirement of providing a good RF field homogeneity over the whole field of view. Results of simulations and experiments showed that a good uniformity can be achieved.

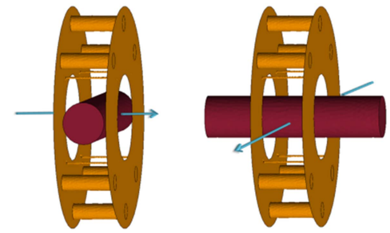


Figure 1: (Left) Phantom placed in the 'in-line' configuration; (Right) Phantom placed in the 'perpendicular' configuration. The arrows represent the axis of the Linac.

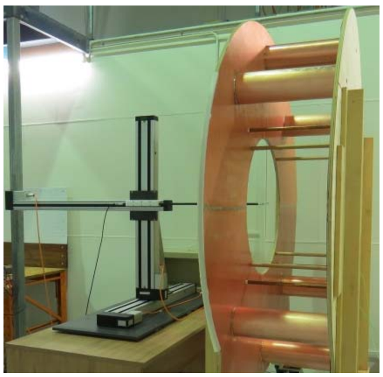


Figure 2: Prototype of the RF coil in the mock-up of the magnet (only the central gap is reproduced). A pick-up coil is fixed on a CNC machine to accurately record the voltage at each position.

METHODOLOGY: As illustrated in Figure 1, access from both sides of the gap had to be provided for either the patient's bed or the LINAC. Furthermore, the rotation of the LINAC around the patient, as required by the conventional radiotherapy, cannot be implemented in the MRI-LINAC if using the in-line configuration. Instead, the patient bed will be rotated or flipped. Thus, the RF coil must be designed to allow for the adequate space to implement this set-up. It was decided to use a modified low pass four leg birdcage coil design, as it can provide a uniform circularly polarised RF field. It is also adapted to the magnet geometry, as the RF rungs can be aligned along the existing mechanical structures which connect the two magnet halves through the central gap, creating no additional obstruction.

The design was first modelled in FEKO (EMSS, SA), where the RF coil was integrated in the magnet gap, represented by its inner flanges (Figure 1). The diameter of the tunnel was 824 mm, and the diameter of the flange was 2100 mm. The six connectors in the gap had a diameter of 150 mm and where placed 895 mm (top and bottom connectors) or 820 mm (diagonal connectors) from the centre of the tunnel. Tubes were used as rungs (inner diameter = 17mm, outer diameter = 19mm, length 480 mm), and were placed 573 mm from the centre of the tunnel of the magnet. A 10 mm gap was left between the end of the rungs and the surface of the magnet. All these materials were modelled as perfectly conductive materials. The capacitors were distributed at both ends of the rungs, and were connected to the flanges. The coil was tuned at 42.5 MHz, and the four legs were excited by 1V voltage sources, with a 90 degrees phase shift between neighbours. The transmit field $B_1^+$was simulated, and its uniformity was observed over a diameter of spherical volume (DSV) of 300 mm.

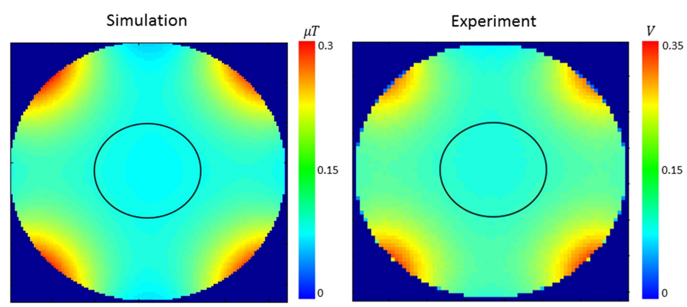
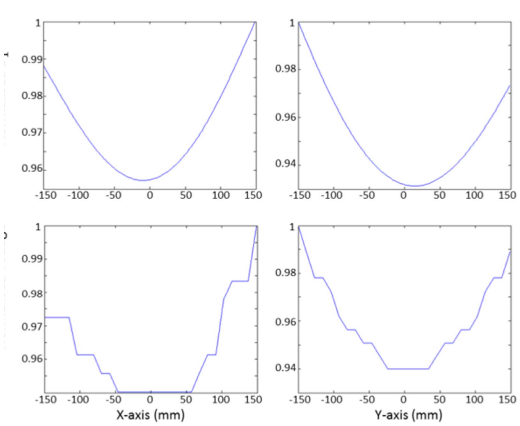


Figure 2: (Left) Result of the electromagnetic simulation; (Right) Recording of the voltage in the mock-up of the magnet; The black circle represents the 300 mm DSV

In order to validate the results of the simulation, a prototype of the RF coil was built, closely resembling the environment modelled in FEKO. A mock-up of the flanges of the magnet was built at a 1:1 scale (Figure 2). The structure was built with timber, and the surfaces in the gap were covered with copper sheets. Because of the skin effect, a small thickness of conductor is enough to model a solid conductor. Copper tubes were used as rungs, and were fixed on the flanges by supports made out of Teflon. Capacitors were connected across the Teflon insulators from the copper tubes to the end-flanges. All dimensions were kept identical to the ones in the simulation. A half-wavelength coaxial cable connected opposite rungs, to simulate a 4-port excitation. After tuning and matching of the coil, a Network Analyser was used to excite each linear polarization mode individually. A pick-up coil was fixed on a 3-axis CNC machine, and connected to an oscilloscope. Our own software was used to control the CNC machine to move the pick-up coil in the centre slice of the gap by steps of 10 mm, and the value of the induced voltage was recorded at each position. Because the voltage is related to the strength of the field (**ref**), it was possible to observe the transmit field. The recording was repeated for each linear polarization mode, and the fields were superimposed. Additional slices along the Z-axis (along the tunnel) were acquired to cover the whole FOV.



RESULTS AND DISCUSSION: In Figure 3, results from the electromagnetic simulation and from the mock-up experiment are compared. Figure (left) shows the simulated B_1^+, and Figure 3 (right) shows the recorded voltages, which are proportional to the transmit field. The 300 mm DSV contour is displayed in both figures. It can be observed that simulation and experiment produce very similar field profiles. The uniformities along the X and Y axes are plotted in Figure 4. As expected, a good homogeneity is achieved over the DSV, with values greater than 95% of the maximum of the DSV. Furthermore, the circular polarization was observed, which will result in an improvement of the SNR by a factor $\sqrt{2}$.

CONCLUSION: In this work, a RF transmit volume coil was designed to work in a MRI-Linac. Because of the novelty of this approach and the specific geometry of the magnet, specific constraints had to be accommodated. To ensure the reliability of the design, the behaviour of the coil was modelled in an electromagnetic simulation, and a prototype was built in a mock-up of the magnet. Results converged to show that the homogeneity of the transmit field in the DSV could be expected to be very uniform. Future work will investigate the behaviour of the coil in the real magnet, to further validate the results. The influence of the load in the different configurations will also be investigated, both in simulations and experiments.

REFERENCES: [1] Liu et al., *Journal of Magnetic Resonance 2012*, 222: 8-15; [2] Keall et al., *Seminars in Radiation Oncology 2014*, 24:203-206; [3] Constantin et al., *Med Phys 2011*, 38:4174-4185;