Bo coil designs for *in vivo* delta relaxation enhanced MR in humans

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Introduction

Targeted MRI contrast agents have recently been developed that have the ability to bind to specific molecules in the body. In some cases, the relaxivity of the agent significantly increases upon binding. For example, MS-325 is an agent that binds to serum albumin in the blood [1,2]. One difficulty is that signal enhancement provided by these 'smart' contrast agents is dependent on the amount of both bound and unbound agent. Delta relaxation enhanced magnetic resonance (dreMR) is a new technology, which allows the specific detection of targeted agents only after they have chemically bound to their target molecule [3]. This method uses an insert electromagnet to modify the main magnetic field as a function of time in an otherwise conventional MR scanner. By interchanging the magnetic field strength during longitudinal recovery of magnetization between two distinct values, the signal provided by the bound agent can be separated from that provided by unbound agent. This is possible by taking advantage of the fact that certain novel MR contrast agents exhibit rapidly changing relaxivity as a function of magnetic field when in the bound state. Previous demonstrations of the dreMR technique have used quite small-bore electromagnet inserts to produce the required shifts in magnetic field. In this study, we investigate the design and performance of insertable electromagnets suitable for performing localized dreMR imaging in human subjects. The two particular anatomical areas of interest are the head/neck, and the prostate; however, this approach may be extended to a variety of other application areas.

Methods

The insert coils were designed using a boundary element (BE) method following the approach of Poole and Bowtell [4]. This method allows the design of coil patterns over virtually any desirable geometry. The first design study was for a head coil. A maximum width of 34 cm in the x-direction was imposed on the coil geometry. The coils were simulated with lengths (along z) of either 30cm or 60cm. In the 30 cm length cases, the target-imaging region was centered 15 cm above the coil surface. Conversely, for the 60 cm length case, the target-imaging region was positioned 15 cm above the coil surface as well as offset 15 cm in the z-direction to allow the positioning of the imaging region to be in the same relative position to an inserted head. The second design study was for a prostate coil. Six geometries were simulated, ranging from completely flat to a half-cylinder (fig.1). The half-cylinder geometry was chosen as a cut off to allow the patient full bore access. These coils were restricted in width to be 52 cm in the x-direction, and allowed a length of 60 cm in the z-direction. The target-imaging region was positioned again 15 cm above the coil surface. In both studies the coils were optimized for maximum efficiency while maintaining a field uniformity of better than 20% over the prescribed spherical imaging regions.

Results

The open head coils ranging in curvature from 180° (half-cylinder) to 324° achieved field efficiencies that increased monotonically from 0.058 to 0.19 mTA^{-1}. The minimum 20% field homogeneity regions were 12 to 15 cm in diameter. The closed head coil case (i.e. 360° curvature) achieved a field efficiency of 0.4 mTA^{-1}, with a 20% field homogeneity of 20 cm diameter. In the prostate coil design study, field efficiencies increased monotonically from 0.03 mTA^{-1} (fully planar design) to 0.093 mTA^{-1} (half-cylinder) while maintaining minimum 20% field homogeneity regions of approximately 12 to 13 cm in diameter. For all designs above, the inductance was held constant at 5000 μH.

Discussion

Distinct contrast using MS-325 has been demonstrated using the dreMR method using field shifts of only 70 mT [5]. The most efficient designs obtained in this study produce shifts of 100 mT using peak currents of: 250 A (closed head coil), 526 A (324° head coil), and 1075 A (half-cylinder prostate coil). These currents are similar to those produced in state-of-the-art gradient amplifier systems today. Because the Bo pulse durations in dreMR sequences are relatively long (on the order of T1 values for the agents of interest), the power-supplies associated with the amplifiers driving the dreMR coil would need to be higher power than typical gradient amplifier systems, and the coils would require significant water cooling. The higher-than-typical inductances for these coils would limit the rate at which they could be switched, but in the dreMR method, very fast switching is not necessary and rise-times of 10-20 ms are acceptable. The 20% field uniformity over the region of interest is acceptable, as the result of field variations in dreMR is a corresponding decrease in contrast-to-noise ratio. Because the field variations would be constant and known, this effect could be included in post-processing. This work demonstrates that with carefully focused electromagnetic design and power considerations not drastically different from current generation gradient systems, dreMR could indeed be conducted in human subjects.

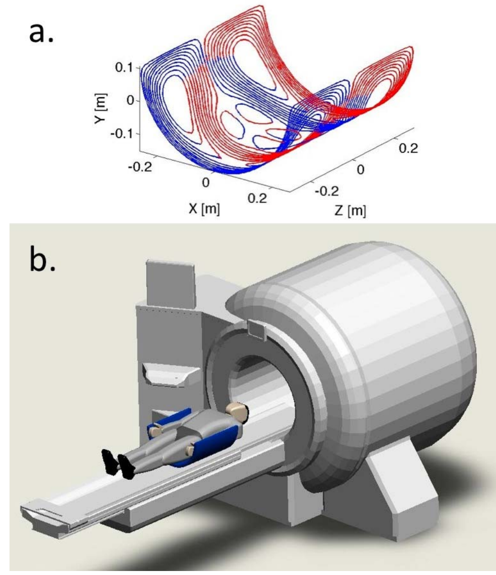

Figure 1. a) The coil pattern for the half cylindrical (180° curvature) geometry open body dreMR electromagnet with maximum width of 52 cm and 60 cm in length. This coil produced a field efficiency of 0.093 mTA^{-1} when scaled for an inductance of 5000 μH. b) Patient lying within insert electromagnet (blue) on MR bed.

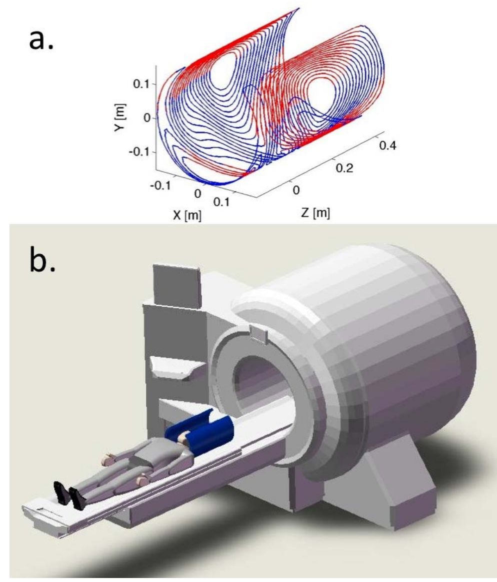

Figure 2. a) The coil pattern for a slightly open geometry head insert electromagnet with a curvature of 288°, radius of 17 cm and 60 cm in length. This coil produced a field efficiency of 0.16 mTA^{-1} when scaled for an inductance of 5000 μH. b) Patient lying on MR bed with head inside coil (blue).

References

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