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Magnetic resonance imaging with RF encoding on curved natural slices

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ABSTRACT

While the idea of using spatial encoding fields (SEM) for image formation has been proven, conventional wisdom still holds that a magnetic resonance imaging (MRI) system begins with a highly uniform magnetic field. In particular, radio frequency (RF) encoding MRIs designed and tested to date have largely used uniform magnetic fields. Here we demonstrate magnetic resonance imaging in a magnetic field with a built-in gradient that gives non-planar slices – curved surfaces – when the nuclear spins are excited with narrow band RF pulses. Image encoding on these naturally occurring non-planar slices was accomplished with RF encoding using a non-linear spatially varying B_1 phase gradient. The imaging methods were demonstrated on a small prototype MRI instrument. The MRI has no switched magnetic field gradients – it is "gradient-free". A low field gradient-free MRI with low mass permanent magnets and simple, low power, RF encoding hardware is ideal for deployment on the International Space Station for the study of astronaut muscle and bone mass loss. Gradient-free natural slice encoding MRI designs would also be portable enough for application in remote terrestrial locations, in emergency rooms and in operating rooms where they can be used with minimally invasive and robotic surgery.

1. Introduction

Ultimately, people who settle on the Moon and Mars will require medical facilities complete with medical imaging capabilities. This includes magnetic resonance imaging (MRI) [1]. To make it feasible for launching into space, these MRIs will need to be considerably lighter than contemporary conventional MRIs, with their massive superconducting magnets and high-power imaging gradients. A previous design for an MRI for the International Space Station (ISS), designed to demonstrate a rocketable MRI, specified a highly homogeneous, wrist-size magnet with linear radio frequency (RF), or TRASE [2], encoding in three directions [3,4]. The design did not have gradient field coils – it was a "gradient-free" MRI design, relying on RF encoding in all three dimensions. In the ISS environment the stray magnetic field needs to be minimized as much as possible without heavy shielding hardware. Therefore a Halbach geometry [5], which ideally produces a uniform field inside the bore perpendicular to the axis, was chosen because it produces a nearly zero stray field outside the bore. To obtain a uniform magnetic field with the Halbach geometry in a small, lightweight volume, a magnet assembly with a 2:1 length:diameter ratio is required [6]. Scaling up such a design from the 50 kg wrist-size ISS-MRI design to whole body size would produce a claustrophobic magnet that would likely have a mass of more than one tonne. Furthermore, with a 2:1 aspect ratio magnet, a head-size system becomes difficult because the shoulders get in the way of accessing the homogeneous field at the center of the magnet.

To overcome the limitations imposed by a 2:1 length:diameter magnet, we designed a Halbach magnet with a non-uniform magnetic field. The magnet was designed with a magnitude gradient perpendicular to the bore of the magnet, in the same vertical direction as the magnetic field. This approach led to a magnet with a 1:2 length:diameter ratio and could likely be pushed to thinner aspect ratios. The gradient was used as a spatial encoding field (SEM) in the vertical direction. The main, B_0, magnetic field produced was composed of foliations of non-planar isomagnetic surfaces that were selected with soft RF pulses. Non-linear RF encoding then was used to accomplish spatial encoding on those natural curved slices.

Various other approaches to designing a viable, portable, gradient-free alternative to the conventional switched gradient field MRI design [7], have been proposed and/or built. Two basic alternative imaging principles are available for gradient-free MRI designs. One alternative principle is to use the gradients present in a generic magnetic field configuration to achieve a SEM image encoding in one direction – a direction that will, in general, be a spatially dependent direction. To obtain another direction of encoding, in this case an angular direction, the main magnet may be rotated to enable two-dimensional imaging [8]. The second alternative imaging principle is to encode images using the RF B_1 field. There are several ways to use B_1 for spatial

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encoding. The original RF encoding method is rotating frame zeugmatography and it works by transmitting RF with a spatially varying B_1 magnitude [9]. Rotating frame zeugmatography, however, requires levels of RF power that are large enough to cause problems with the specific absorption rate (SAR) and tissue heating [2]. The Transmit Array Spatial Encoding (TRASE) method imposes, instead, a linearly varying spatial phase gradient on the transmitted B_1 field. With a linearly varying B_1 spatial phase gradient, the image may be reconstructed using a Fourier transform. A non-linear spatial phase gradient, associated with a more generic B_1 field, may be used with more general image reconstruction algorithms [10,11]. Finally, B_1 receive fields, as opposed to the transmit fields discussed so far, will naturally have a spatially varying magnitude that can be employed to achieve spatial encoding. Spatially varying B_1 magnitude is exploited to speed up conventional MRI data acquisition with the parallel imaging methods of SENSE [12], which combine various datasets in the spatial domain, and SMASH/GRAPPA [13,14], which combine data in the spatial frequency *k*-space domain. By mechanically or electronically rotating the receive coils, it may be possible to achieve two-dimensional image encoding using the spatial variation of the receive B_1 field alone [15].

In this work we designed an MRI to combine SEM encoding in one direction with non-linear RF phase encoding in the other two directions. The SEM encoding here is essentially non-planar slice selection – natural slice encoding. Our work differs from others as follows. Earlier work on linear RF-encoding (TRASE) was performed with TRASE encoding in two directions and slice selection in the third direction using a conventional magnetic field gradient [2]. Our prototype uses non-linear RF encoding in two directions and slice selection in the third direction on fixed, curved, natural slices. The MGH/MIT [8] group used a built-in non-linear SEM field in the one direction and a rotating magnet to resolve a second direction. The original MGH/MIT prototype did not spatially encode NMR data in a third direction but TRASE encoding has recently been added along the bore of the magnet [16]. We have built another prototype in our laboratory that uses B_1 spatial encoding by the receive coils in the radial direction with a motorized rotating magnet to achieve angular encoding [17]. There are other approaches to portable MRI (e.g. a head scanner based on electromagnets [18]) that have been published but they are not gradient-free designs.

This paper is organized as follows. In Section 2 the design and construction of a prototype MRI, built to demonstrate the new imaging approach, is described. In the methods section, Section 3, the imaging RF pulse sequences and image reconstruction methods are described. In Section 4 the images made with our prototype are presented. Possible avenues for future improvements are discussed in Section 5 and the paper concludes with Section 6.

2. Hardware design and construction

A “wrist size” prototype MRI was built to verify the approach of using non-linear RF encoding on natural slices in a non-uniform B_0 field. Owning to various factors, discussed later, the scan time was about 50 h – so the imaging capability of the instrument was assessed with fruit and vegetable slices instead of with volunteer wrists. The required hardware for the imaging approach used is very simple: a magnet and an RF system. The design of the RF transmit system we used is unique in its hardware simplicity and relies on image post-processing to compensate for the non-linear spatial B_1 phase variation it produces.

2.1. Magnet

The magnet was designed around a Halbach geometry [5] and was constructed with 60 1 in. × 1 in. × 1/2 in. thick NdFeB, Grade N52, magnet blocks (part number BX0 × 08-N52, K&J Magnetics, Inc., Pipersville, PA, USA). This choice represents the largest NbFeB cubes that we believed we could safely handle. Magnet blocks were organized into pairs, allowing for a thin yet strong wall between the two magnet blocks. A 3 mm wall proved thick enough to have the required structural strength, yet thin enough so that the magnet blocks are very much held in place by mutual attraction.

To determine magnet orientation, an iterative-optimization algorithm was used that sets the orientation of each magnet-pair in the structure to best match the desired field profile. Usually, a uniform magnetic field is desired over the imaging field of view. However, as we plan on using a fixed gradient in the up/down direction, we selected a linearly increasing field profile, starting at 490 G (2.12 MHz) at the bottom, rising to 500 G (2.16 MHz) at the top of the bore. We used a gradient descent optimization algorithm, using a least squares error function. Depending on the number of iterations used, the run time for the algorithm is anywhere from 20 s to 30 min. This approach allows for both quick outlining of the design as well as thorough optimization to ensure good performance of the final design. The quick outlining approach was primarily used to determine how many magnet blocks to use and how to space them amongst the overall structure. Experimentation suggested a simple co-centric design with three rings of magnet blocks (10 pairs each) spaced evenly along the cylindrical structure. As the RF transmit coils occupy primarily the central region, we were able to angle-in the front and back magnetic rings, which increased available magnetic field slightly.

While the Halbach configuration has been used in a variety of studies [5,6,8], it is designed to produce a uniform magnetic field, not the varying field we seek. The magnet as constructed here and shown in

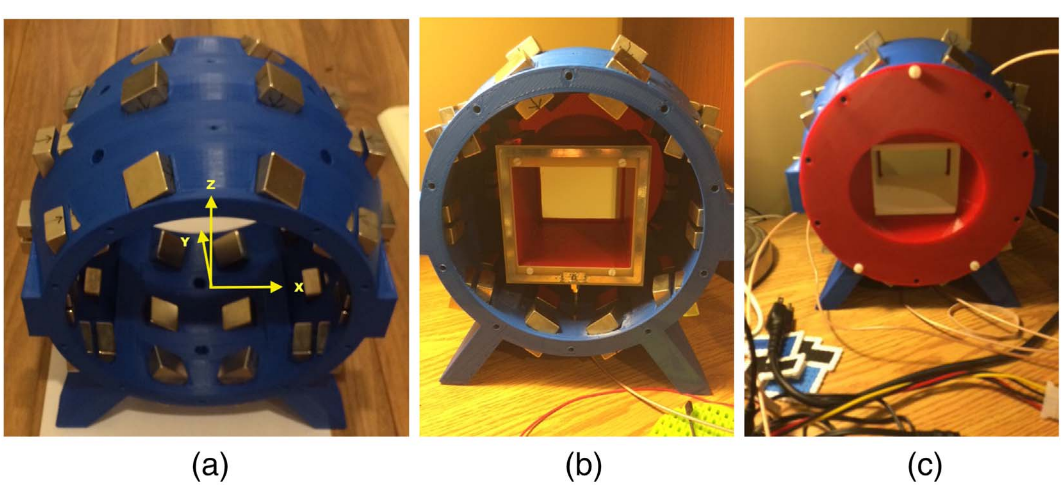

Fig. 1. Magnet assembly. (a) Assembly showing component permanent magnet blocks. (b) Cube transmitter integrated into the magnet. (c) Completed magnet assembly. The outer diameter of the front bore is 207 mm, while the inner diameter is 177 mm.