2D Imaging in a Lightweight Portable MRI Scanner without Gradient Coils

Clarissa Zimmerman Cooley[1,2], Jason P. Stockmann[1,3], Brandon D. Armstrong[1,3], Mathieu Sarracanie[1,3], Michael H. Lev[4,5], Matthew S. Rosen[1,3], and Lawrence L. Wald[1,5]

[1]*A. A. Martinos Center for Biomedical Imaging, Dept. of Radiology, Massachusetts General Hospital, Charlestown, MA, United States,* [2]*Dept. of Electrical Engineering and Computer Science, Massachusetts Institute of Technology, Cambridge, MA, United States,* [3]*Dept. of Physics, Harvard University, Cambridge, MA, United States,* [4]*Dept. of Radiology, Massachusetts General Hospital, Boston, MA, United States,* [5]*Harvard Medical School, Boston, MA, United States*

PURPOSE: As the premiere modality for brain imaging, MRI could find wider applicability if lightweight, portable systems were available for siting in unconventional locations such as ICUs, physician offices, ambulances, ERs, sports facilities, or rural healthcare sites.

METHODS: A small rotating permanent magnet array with an inhomogeneous field is used to create a portable MRI scanner. The field distribution is used for image encoding instead of gradient coils. This reduces the weight and acoustic noise of the scanner, and allows it to be run from a standard power outlet.

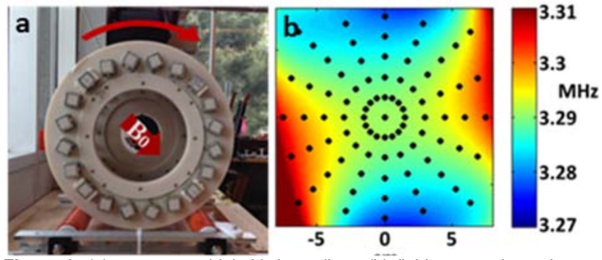

Figure 1: (a) magnet on high friction rollers, (b) field map estimated as a polynomial fit to the measured field probe points (black dots).

The previously described 45 kg (36 cm dia. 35.6 cm length) Halbach cylinder magnet[1,2] (Fig. 1) uses 20 rungs of NdFeB magnets to create a 77.3 mT field (3.29 MHz proton freq.) with 32 kHz variation in the center slice 16 cm dia. FOV (mapped using small field probes). As the magnet is physically rotated around the object using a stepper motor, a RARE-type spin echo sequence (40 kHz, 256pt readout) records the generalized projections onto the nonlinear field pattern. The individual echoes are averaged to improve SNR. A 20cm diameter solenoid with a low Q (~60) is used to excite the bandwidth of spins using short (50 μs) pulses.

The non-bijectivity of the approximately quadrupolar spatial encoding magnetic field (SEM) causes image aliasing that we disambiguate using parallel imaging as described by Schultz *et al.*[3] We use an 8-channel receive array (Fig. 2a) with 8 cm circular loops for this purpose. The coil sensitivity map (Fig. 2b) is modeled using the Biot-Savart law and is different for each magnet rotation angle. The coil sensitivity profiles and field maps at each rotation angle are used to form the encoding matrix. We solve for the image using the Algebraic Reconstruction Technique (ART)[2,4].

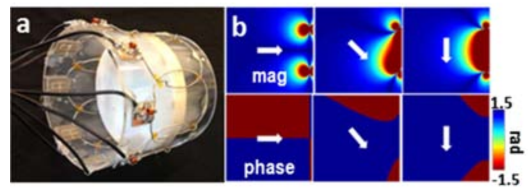

Figure 2: (a) 8-channel RX array coil with disk phantom. (b) Calculated B_1^- profiles in the center plane of a surface coil located at the right side of the FOV. The arrows show three representative orientations of B_0.

RESULTS: Figure 3 shows simulated images based on a 3T reference brain image and a 2.5 mm grid as the "object." The SEM field map is rotated (181 angles spaced 1°) to generate model data to which noise is added to match experimental data. A simulated image is then reconstructed with ART. Fig. 3b shows the simulated image expected from the measured SEM. Fig. 3c shows the image when a linear component (of 500 Hz/cm) is added to the measured SEM to reduce the "encoding hole" in the center. Figure 4 shows experimental images of a "MIT/MGH" phantom ($CuSO_4$-doped water, 1.7 cm thick, 13 cm dia.) acquired with 7 Rx coils, 32 averages of a 6 spin-echo train (TR = 550 ms, echo spacing = 8 ms) using 91 magnet rotations of 2°. Data is also acquired from a single field probe at each rotation to track field drift for use during reconstruction (Fig. 4b). The lengthy acquisition time of 66 minutes results from serially acquiring data from the 8 coils into a single channel on the console and would be reduced to 7.3 minutes with true parallel acquisition. Figure 4c shows a 1cm thick lemon slice imaged using 5 Rx coils, 128 echo train (TR = 4500 ms, echo spacing = 8 ms), and 181 magnet rotations of 1°. The total acquisition time was 93 minutes (15.5 minutes w/ parallel Rx).

DISCUSSION: The nonlinear SEM of the Halbach magnet results in variable resolution over the FOV, including notable blurring in the center where the encoding field is spatially uniform (Fig. 3b,d and 4c). If a sufficiently-strong linear term was added to the SEM, the spatially uniform encoding field region would not coincide with the axis of rotation. In this case, (simulated in Fig. 3c) the "blind-spot" moves around the object resulting in less severe blurring. The simulations in Figure 3 show the theoretical resolution of the scanner when systematic errors are eliminated. These errors most likely result from inaccuracies in the field map and coil sensitivity profiles propagating through the iterative reconstruction[5]. Field map errors from temperature drifts are significant but can be measured and included in the encoding matrix and mitigated in the reconstructed image (Fig. 4b).

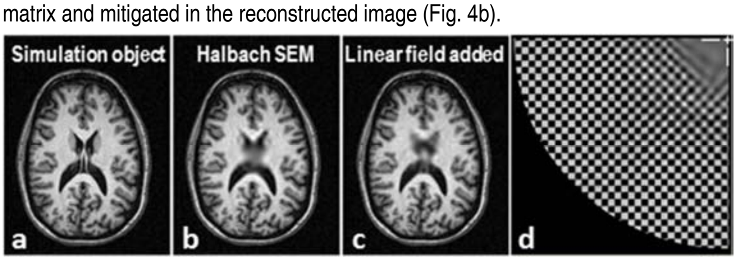

Figure 3: Simulations. (a) T1-weighted brain reference. (b) Simulated reconstruction using measured field. (c) Simulated with measured field + linear field. (d) 2.5 mm checkerboard simulated reconstruction. Only 1 quadrant shown, center of FOV marked in upper right.

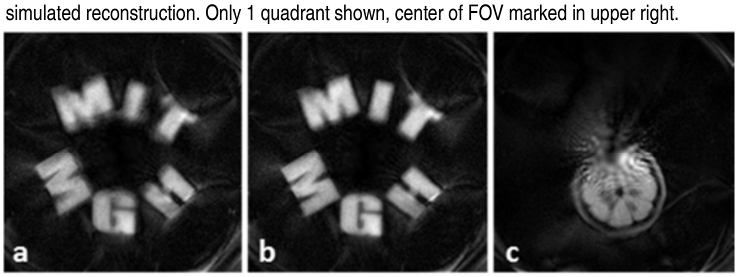

Figure 4: Experimental 16 cm FOV images. (a,b) 1.5cm thick phantom filled with $CuSO_4$-doped water. (a) Temperature drift not corrected, (b) temperature drift corrected. (c) 1 cm thick lemon slice.

CONCLUSION: Using an inhomogeneous rotating magnet for spatial encoding in lieu of gradient coils, we have constructed and demonstrated a lightweight scanner for 2D MR imaging with minimal power requirements. The 2D proof-of-concept images from this nearly head-sized imager show the ability of this encoding scheme to produce sufficient spatial resolution and sensitivity for the detection and characterization of many common neurological disorders such as hydrocephalus and traumatic space-occupying hemorrhages. Future work perfecting the calibration methods is likely to bring experimental image quality closer to the theoretical limit, but the resolution of the current system is sufficient for identifying gross pathologies.

The addition of encoding along the 3rd axis (the axis of the cylinder) is an obvious requirement for medical applications. Transmit Array Spatial Encoding (TRASE)[6,7] is a promising method for achieving this. With the future implementation of true parallel imaging and 3D encoding, this scanner has the potential to enable a truly portable, low-cost brain imaging device.

REFERENCES: (1) Zimmerman C, ISMRM 2012. (2) Cooley CZ, ISMRM 2013. (3) Schultz G, MRM 2010. (4) Gordon R, J. of Theoretical Bio. 1970. (5) Stockmann JP, MRM 2013. (6) Sharp JC, MRM 2010. (7) Sharp JC, NMR in Biomed. 2013.

ACKNOWLEDGMENTS: The authors thank C Lapierre, M Christensen, B Guerin, and S Cauley. Support by DoD/USAMRRA W81XWH-11-2-0076 (DM09094), NIH/NBIB *P41EB015896,* NIH *T90DA022759/R90DA023427.*