Development of a novel 1 mT planar B-zero coil for patient respiratory motion compensation in magnetic resonance imaging

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Introduction

A novel B_0-coil has been developed to reduce respiratory motion artefacts in MRI. Combined with a local MEMS detector microcoil, deployed in an endoscope for MR imaging of cholangiocarcinomas, images with sub-millimetre resolution can be achieved. The B_0-coil has been designed to generate 1 mT field, parallel to the main field of a MRI scanner, with 80 A_{pk} and to be homogenous to within ± 1% in a 150 mm DSV. We detail the design, construction and testing of the B_0-coil.

Tissue movement has always been a problem with MR imaging of the abdomen, and particularly with imaging of the liver. Modern approaches to this problem have tended to concentrate on fast imaging methods such as EPI (1), breath-holding (2) or RARE (3) and more recently parallel imaging methods such as SENSE (4) or SMASH (5). However, none of these approaches is relevant during high resolution MR guided endoscopy. In the study for which the system described here was developed, the target is the detection of cholangiocarcinomas, which are a tumor of the bile duct, that need to be treated before they become more than about a millimetre in size. The tumor incidence is rapidly increasing in western populations (6), as it is a consequence of liver disease, resulting from obesity and alcoholism. It has been found that even liver transplantation is followed by recurrence if the tumor is allowed to grow much above the millimetre size. Image resolutions of 0.5mm are the minimum needed. In this project we use the biopsy channel of the duodenoscope to provide a route for a catheter mounted micro-electro-mechanical systems (MEMS) based microcoil (7) detector to be placed immediately adjacent to the likely location of the lesion inside the bile duct. Sections of one of the lumens of the catheter supporting the microcoil can be filled with an MR-visible material such as a solution of a gadolinium contrast agent so that the microcoil can then act as its own fiducial marker as well as having the capability of imaging tissues immediately outside it. Alternatively, landmarks such as the diaphragm can be tracked using Navigator Sequences (8). Images from a small field of view (FOV) inside tissue which is capable of moving distances large compared with the range of the detector raises is a major problem. It is the purpose of this paper to describe an approach which minimises the effect of patient respiratory motion using a B_0-coil.

Motion correction in all three imaging directions is essential. Correction along the readout, phase encoding and slice select gradient directions can readily be performed (by shifting demodulation frequency or using a method such as ROPE (9) or dynamically adjusting RF excitation frequency) but require complex sequences and the ability to intervene in real-time in a controlled manner. Other methods (10,11) involve collecting additional data, and retrospectively averaging or ordering them. In order to avoid these problems an alternative approach can instead be implemented if the scanner magnetic field B_0, is adjusted by a small ΔB_0, as the sequence proceeds so that at any time the microcoil appears to be at the same place in the space defined by the gradient fields as it was at the start. Hence the field is adjusted by $\Delta B_0 = G_S.\Delta_S$, where G_S is the slice select gradient and Δ_S is the displacement to be corrected perpendicular to the slice plane. The data thus will be consistent at all times, and motion effects are eliminated from it. The insert coil produces a homogeneous field parallel to B_0 and can be switched with the same response as the gradient coils. For a typical 1.5 T magnet with a homogeneity of 10 part-per-million and a slice select gradient of 10 mT/m this is equivalent to generating a positional error of less than 1 mm for an insert coil with 1 % field inhomogeneity in the DSV.

B_0-Coil Design and Construction

Figure 1 (left) shows one quadrant of the planar coil with the number of turns, dimensions in millimetres, and the direction of current flow needed to generate magnetic field parallel to B_0. The coil length is 37 m and has a DC resistance of 0.36 Ω and an inductance of 140 μH at 200 kHz. The coil has symmetries along the dotted lines shown in the figure. Centre of figure 1 shows the final coil assembly with dimensions of 775 mm (L) x 475 mm (W) x 300 mm (H). The coil weight is about 10.8 kg.

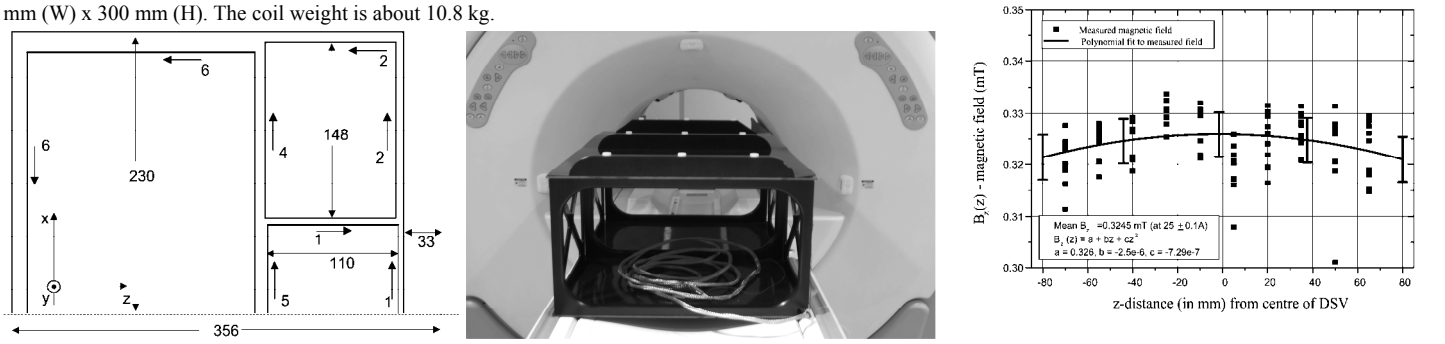

Fig. 1. Design of the B_0-Coil (left), final assembled coil in a MRI scanner (centre) and the field distribution measurements in the DSV (right).

Results

To map the magnetic field within the DSV the field was measured using a Gaussmeter (Hirst GM08, Cornwall, UK) with an axial Hall probe with a quoted accuracy of ± 0.35% traceable to national standards. Figure 1 (right) shows measurements of the field in the DSV to be homogeneous with a mean value of 324.5 ± 4.3 μT (1σ) at 25 A, which also agreed to better than 2 % with the calculated value using the Biot-Savart law. MRI experiments on a test phantom confirmed axial and sagittal image slice shifts of approximately 12 mm and 3 mm, respectively, using only 5.5 A_{pk} in the B_0-coil supplied during slice select for either axial or sagittal imaging. These results clearly demonstrate the principle of image slice shifting by modulating the B_0-field of a MRI magnet using a relatively small field generated in an insert coil.

Acknowledgements

Funding from EPSRC, under Grant EP/E005888/1 "In-vivo biliary imaging and tissue sampling", is gratefully acknowledged and Prof R. Syms, Prof W.M.W.Gedroyc, Prof. S. D.Taylor-Robinson, Dr M. M. Ahmad, Dr Shahid Khan, Prof. Mark Thursz, Dr C.Wadsworth and Mr M. Rea for useful discussions.

References

1. Mansfield P. Multi-planar image formation using NMR spin echoes. J. Phys. C, Solid State Physics, 1977, 10; L55 – 58.
2. Haase A. Snapshot FLASH MRI: Applications to T_1, T_2 and chemical shift imaging. Magn. Reson. Med., 1990, 13; 963 – 969.
3. Hennig J, Friedburg H. Clinical Applications and Methodological Developments of the Rare Technique. Magn. Reson. Imaging, 1988, 6; 391 – 395.
4. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI, 1999, 42: 952 – 962.
5. Sodickson DK. Tailored SMASH image reconstructions for robust in vivo MR imaging. Magn. Reson. Med., 2000, 44; 243 – 251.
6. Taylor-Robinson SD, Foster GR, Arora S, Hargreaves S, Thomas HC. Increase in primary liver cancer in the UK, 1979-94. Lancet. 1997, 350:1142 – 1143.
7. Ahmad M. M., Syms R. R. A., Young I. R., Mathew B., Casperz W., Taylor-Robinson S. D., Wadsworth C. A., Gedroyc W. M. W. Submitted to J. Micromech. Microeng.
8. Firmin D and Keegan J, Navigator echoes in cardiac magnetic resonance, J. Cardiovascular Magnetic Resonance, 2001, 3 (3); 183 – 193.
9. Bailes DR, Gilderdale DJ, Bydder GM, Collins AG, Firmin DR, Respiratory Ordered Phase Encoding (ROPE), a method for reducing respiratory motion artefact in magnetic resonance imaging. J. Comp. Assist. Tomogr., 1985, 9; 835 – 838.
10. Haacke EM, Patrick JL. Reducing motion artefacts in two-dimensional Fourier transform imaging, Magn. Reson.Imaging, 1986, 4: 359 – 376.
11. Stark DD, Hendrick RE, Hahn PF, Ferrucci JT. Motion artefact reduction with fast spin-echo imaging. Radiology, 1986, 164; 183 – 191.