Construction of a 5000 G/cm Z-Gradient Coil for q-Space Microscopy

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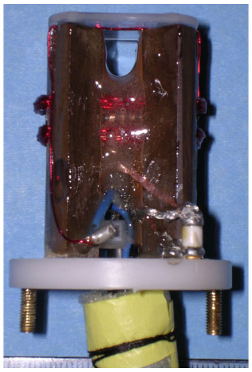
Introduction

Q-space imaging (1) has the potential to provide detailed quantitative information on the geometry of structures at cellular resolution (2, 3). However, the order of magnitude of the size of restrictions that can be probed hinges on the achievable gradient amplitude. Further, since q_{max} determines the resolution ($\Delta r=1/q_{max}$) in the displacement domain, and imposing the condition $\delta<<\Delta$ (the gradient pulse duration should be short relative to the diffusion time), this puts very high demands on gradient performance. For example, for a displacement resolution of 1 µm and δ=1 ms, G_{max} = 2,300 G/cm would be required ($q=\gamma\delta\square$), which is more than one order of magnitude greater than gradient strengths typically available. In this work, we constructed a high-amplitude (5000 G/cm) pulsed field z-gradient coil (micro-z) with integrated solenoidal RF coil, and combined it with the x- and y-gradients (100 G/cm) of a vertical-bore 400 MHz NMR microimaging system. Preliminary q-space microscopy experiments on 4.5 µm diameter polystyrene microspheres showed the expected q-space diffraction peak.

Description of Hardware

Following the design of Callaghan, et al. (4), the supporting structure for the gradient/RF coil assembly (see photos) was made to plug into the RF insert of a Bruker DMX 400 micro-imaging system (Micro2.5 gradients with BAFPA40 amplifiers). In order to create a vertical (z-axis) quadrupolar gradient field, enamel-coated magnet wire (0.36 mm dia.) was wound in four horizontal bunches of 23 strands each, resulting in a coil inductance of 21 µH and a resistance of 1.7 Ω. A horizontal glass NMR tube (o.d.=4.0 mm) was centered between the wires, in addition to a smaller removable glass NMR sample tube (i.d.=2.4 mm). On the outside of the 4 mm tube a 3-turn solenoidal RF coil was wound from copper foil (13.4 µs 90° square pulse at 50 W), with variable tuning/matching capacitors mounted in the base. To provide mechanical stability and restrict Lorentz force vibrations, the entire gradient/RF coil assembly was potted in epoxy. In addition, the assembly was equipped with a plastic cover that made an O-ring seal against the walls of the bore and provided set screws to clamp the micro-z coil.



Results and Discussion

The micro-z homogeneity was estimated by calculating the gradient field amplitude from this arrangement of wires using the Biot-Savart law (Fig. 1). The gradient field is non-uniform close to the wires at the corners of the grid, but provides sufficient uniformity over the sample region (~ 1.26 T/m/A ± 2%). Good uniformity also was demonstrated by an undistorted image of a water-filled tube, obtained using the micro-z for readout and the vendor's x and y coils for phase-encoding and slice selection. The micro-z linearity with applied current was assessed from a capillary profile projected onto the read axis, yielding a slope of 1.26 T/m/A which agreed with calculations and corresponded to 50 T/m at peak amplifier output of 40 A. Eddy current effects were minimal: rise time of the micro-z pulses was comparable to that of the vendor's gradients (~100 µs), and only minor pre-emphasis adjustment was required even for 40 A applied on the z channel.

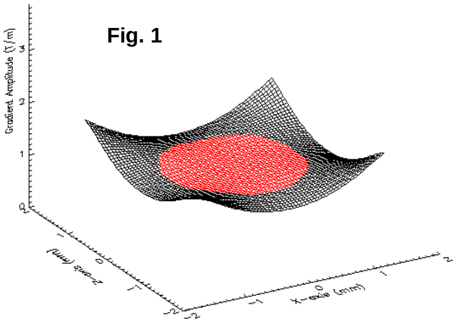


Fig. 1

The potential of the gradient/RF coil assembly is illustrated with a q-space experiment performed on a preparation of closely packed polystyrene microspheres of 4.5 µm diameter (Duke Scientific, USA). The results are shown in Fig. 2, where a diffraction maximum is observed at a *q* value corresponding to 1/*a* (*a* = 4.5 µm) thus demonstrating the capability of the micro-z coil to generate gradient amplitudes significantly larger than those accessible by the commercial gradient set. Furthermore, these data represent *q*-space diffraction phenomena at a length scale smaller than that of any previously published study, being on the order of the length scale of cells.

In conclusion, a relatively low-cost, high-amplitude quadrupolar gradient coil was constructed and integrated with an existing commercial microimaging system, providing new capabilities for localized q-space microscopy. The intended goal is to achieve a resolution in the displacement domain of about 1µm, requiring gradient performance within the range for which this coil has been designed.

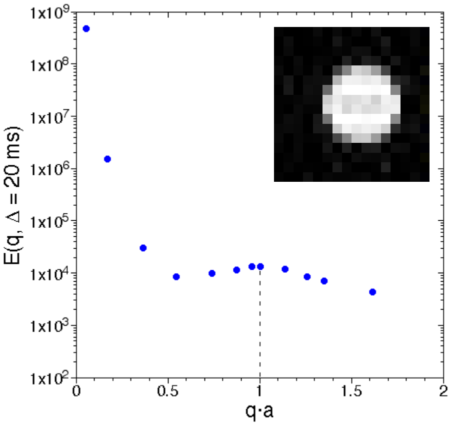


Fig. 2 Plot of image intensity (echo) attenuation versus the product *qa* from a sample of packed polystyrene micro-spheres of diameter *a* = 4.5 µm in water in a 2.4 mm diameter tube obtained at 400 MHz with Δ = 20 ms. Note the diffraction peak occurring at the predicted *q* value (gradient amplitude = 175 G/cm) where *q* = 1/*a*, in excellent agreement with actual bead diameter.
Inset: low-resolution diffusion-weighted spin-echo image of the sample (voxel=313x313x4000 μm^3, matrix=16x16, FOV=5x5 mm^2, TR=2 s, TE=25.8 ms, Δ=20 ms, δ=3 ms, diffusion gradient direction = read axis (vertical axis in the image), NEX=25, scan time=13 min 20 s).

Acknowledgements

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References

1. Callaghan, PT, Principles of Nuclear Magnetic Resonance Microscopy (Oxford, 1991). 2. Cory, DG and Garroway, AN, *Magn. Reson. Med.* **14**, 435-444 (1990). 3. Assaf, Y, et al., *Magn. Reson. Med.* **44**, 713-722 (2000). 4. Callaghan, PT, et al., *J. Magn. Reson.* **133**, 177-182 (1998).