



## Progress in high field MRI at the University of Florida

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### Abstract

In this article we report on progress in high magnetic field MRI at the University of Florida in support of our new 750MHz wide bore and 11.7T/40cm MR instruments. The primary emphasis is on the associated rf technology required, particularly high frequency volume and phased array coils. Preliminary imaging results at 750MHz are presented. Our results imply that the pursuit of even higher fields seems warranted. © 2002 Elsevier Science B.V. All rights reserved.

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### 1. Introduction

The development of higher field magnets for MR imaging and spectroscopy continues as strongly as ever despite the increased cost and associated technological difficulties. The primary drive remains the expectation of increased signal-to-noise ratio (SNR) and spectral dispersion that can be achieved at higher fields [1,2]. In particular, the increased SNR can be traded for decreased data acquisition time and/or facilitate data acquisition from lower concentrations and smaller samples in spectroscopy or from smaller voxels in imaging.

As part of this continuing evolution, the University of Florida (UF) has purchased the world's first 11.7 T/40 cm MRI/S instrument (a Magnex Scientific magnet with a Bruker Instruments console) and a 750 MHz/89 mm vertical Bruker Instruments system (the second in the world and the first with imaging capability, though other systems are now coming online). Both

instruments are shown in Fig. 1. These magnets operate at the highest magnetic fields (frequency) for their bore size and bring with them considerable challenges in engineering in the form of both magnet technology and ancillary hardware, particularly the radio frequency (RF) coils. The RF coil technology associated with the instrument is particularly important, because there would be little point in increasing the magnetic field strength to improve SNR if that SNR gain then was lost through RF coil limitations.

In the spirit of the ISMRM workshop on MR hardware at which this work was presented, this paper summarizes our recent efforts to develop suitable RF coil technology in support of these new instruments. Although both instruments are intended for spectroscopic applications as well, the focus here will be on MR imaging. Preliminary images on the new instruments are presented.

### 2. The magnets

The Advanced Magnetic Resonance Imaging and Spectroscopy (AMRIS) facility in the McKnight Brain

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Institute (MBI) at UF house five magnets. It is supported by initial funding primarily from the Department of Defense and Department of Veterans Affairs and is in partnership with the National High Magnetic Field Laboratory (NHMFL), which supports an outside users program for the facility. With the decision to build the facility and following a discussion meeting on the merits of higher fields at the NHMFL [3], it was decided to purchase novel high field systems that pushed the state of the art. The first magnet to be constructed was a 40 cm bore at the highest field that could be reasonably achieved so that small animals (including primates) could be accommodated. Initially a 9.4 T magnet was envisaged, but it was soon realized that a pumped helium system operating at a reduced temperature ( $\sim 2.2$  K) with conventional superconducting wire technology could potentially achieve 11.7 T. In just 11 months, Magnex Instruments Inc. (UK) constructed the magnet, which reached approximately 10.4 T in the factory. In a subsequent test, the magnet was moved into the AMRIS facility (see Fig. 1) and made 11.7 T on the first attempt without a training quench—quite an achievement. The magnet is inside a shielded room, the shielding consisting of four walls composed of 180 tons of steel. The magnet has since been interfaced with a Bruker Instruments Ltd. console.

Unfortunately, before the rest of the hardware was successfully installed and tested, a leak in the cryogenics occurred and the subsequent development of ice blockages has resulted in the instrument being powered down for repairs. At the time of writing, the ice appears to have been cleared and the system is being re-cooled. In the meantime, a sister magnet at 11.7 T with a slightly smaller bore (33 cm) has been constructed by Magnex and installed at the NIH. We have kindly been pro-

vided access to this instrument until ours is functional (see below).

With present technologies, the construction of even higher field magnets necessitates a reduced bore size. Thus the second instrument purchased for the AMRIS facility was the highest available wide bore (89 mm) magnet, one of a batch of the first three constructed by Bruker Instruments. A wide bore was chosen over a conventional narrow bore (52 mm clear bore) so that mice could be accommodated. The first of these magnets was a solid-state instrument, with the second (for UF) intended to be able to perform both imaging and high-resolution spectroscopy. The 750 MHz/89 mm vertical bore instrument also is a pumped helium system (Fig. 1) and has a Bruker console. This instrument was installed and successfully made field on the second attempt. This instrument has performed well for imaging over the last year, and preliminary data is presented below.

Initial attempts at high-resolution spectroscopy on the 750 MHz/89 mm suffered from instabilities arising from eddy currents generated when the wide bore room temperature shims were disconnected and the high-resolution shim set connected. It appears that the wide bore shims coupled strongly enough with the magnet cryoshims to cause instabilities that last a few days, making the switch from imaging to high resolution clearly impractical. Bruker resolved this problem by installing a second separate shim power supply train for the wide bore shims so that they did not need to be disconnected but remained powered. Since then, high-resolution spectroscopy has met specifications.

### 3. RF coil developments

As described in Section 1, the SNR advantages obtained through judicious RF coil designs at lower main magnetic field strengths must be available at higher fields if the full advantage of the increased field is to be realized. Thus, at higher fields, good quality volume coils must be constructed. Additionally, phased array coils have become a prominent technology in MRI for improving the SNR in images, and it is desirable that this technology also be implemented on the new magnets. Several groups are working in both areas and we have explored these requirements for our new magnet systems.

#### 3.1. High frequency, large volume RF coils

As the required resonance frequency of relatively large volume coils (1–30 cm diameter) increases, the RF wavelength approaches a significant fraction of the physical size of the coil, making tuning difficult if not impractical. The mainstay of relatively high field MR is



Fig. 1. Photographs of the installed magnets, both of which are pumped magnets. Left, 11.7 T (500 MHz)/40 cm; right, 750 MHz (17.6 T)/89 mm.

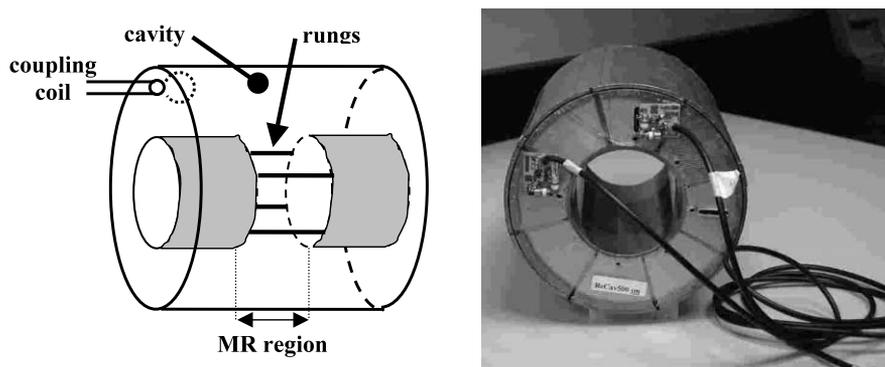


Fig. 2. Schematic of the re-entrant cavity (ReCav) coil and a photograph of a finished coil. Note that the numbers of rungs in the MR region is variable with the required frequency (we have used four or eight splits with capacitors) and that one (linear) or two (quadrature) coupling coils may be used. The outer tube, end plates and re-entrant cylindrical section are one piece of copper forming the shield. This shield is split to reduce eddy currents.

the birdcage coil [4], which achieves higher frequencies than conventional coils through the parallel nature of its design and thus reducing the inductance so that the achievable resonance frequency is increased. The birdcage, however, has its limits and has led to the development of alternative geometries, in particular the cavity resonators [5,6] that have been pioneered for MR applications primarily by Vaughan et al. [7]. We have been developing a modified version of the cavity

coil for high field MR that attempts to minimize far field radiation losses by having a section of the cavity re-enter the inner coil volume (see Fig. 2). Hence, we have named this a re-entrant cavity coil (ReCav) [8]. To date, an equivalent birdcage coil geometry has proven as effective as the ReCav at 4.7 T (both coils having 9.5 cm internal diameter). Initial studies at 9.4 T demonstrated an advantage of the ReCav over the birdcage with a 46% better SNR on a loading phan-

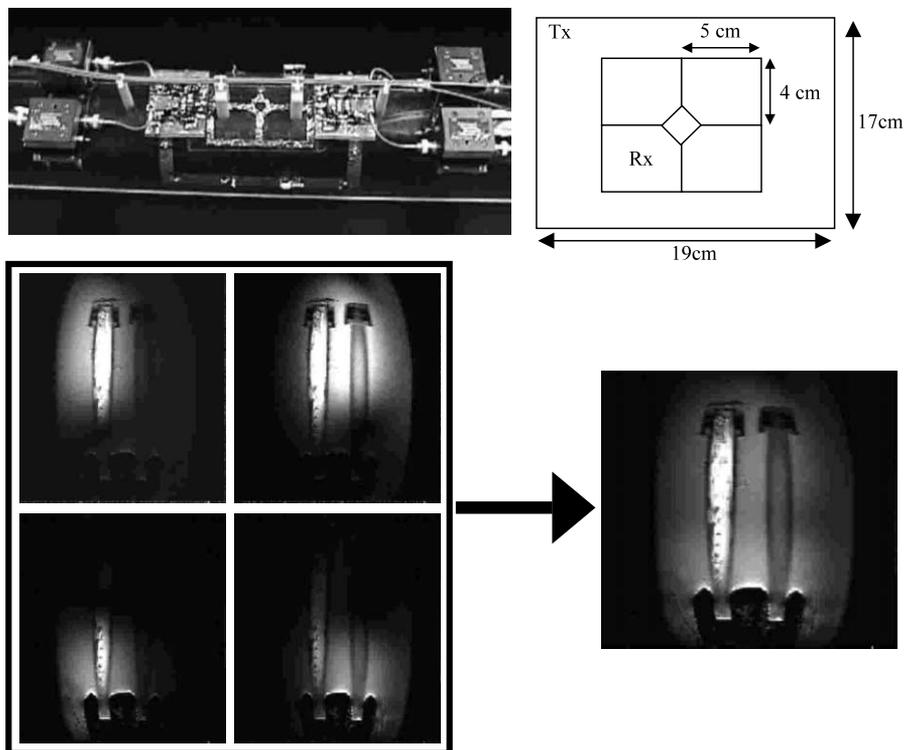


Fig. 3. Top left: A 200 MHz phased array coil with the central array showing the four-coil array (Rx) around which can be seen the larger single loop transmit coil (Tx). The four boxes on the coil (two left, two right) are the preamplifiers that are now mounted more conveniently at the back of the magnet. The rest of the circuitry is for tuning and decoupling of the coils. The dimensions of these coils are shown in the accompanying schematic (top right; note that the coils are laid out on a 14 cm diameter acrylic tube). Bottom left: four separate images from a phantom can be combined to give a composite phased array image (bottom right).

tom. We presently are evaluating similar coils at 11.7 T, through access to the 11.7 T/33 cm instrument at the NIH courtesy of Drs Koretsky and Silva, and have obtained results indicating an improved performance of the ReCav over the birdcage. Preliminary results on phantoms are encouraging, and we expect to be able to construct fully function cavity volume coils at these high frequencies.

Additionally, we are modeling the RF characteristics of these resonators with our co-authors in Queensland, Australia and Hershey, PA, USA. These models show the ReCav coils to behave as a true cavity. These simulations, which will include the influence of coil loading, will be used to explore the effectiveness of modifications in the coil geometry in order to optimize its MR characteristics. At this point, we are confident that large volume, high-frequency coils can be constructed at 11.7 T, making this instrument truly effective for small animal imaging.

### 3.2. Phased array coils

Phased array coils offer improved SNR over equivalent single coils and have become important for efficient MR imaging [9], with potential application for spec-troscopy. Recently, we constructed a transceive

pelvic phased array RF coil for our 3 T system, which was used to obtain images and  $^1\text{H}$  spectra of the human prostate [10]. Our long-term goal is to extend these technologies to higher frequencies for our new instruments. In these regards, we have recently constructed a receive-only four-coil phased array with a separate transmit coil [11,12] that operates at 200 MHz (4.7 T). The coil is shown in Fig. 3, along with example images of the functioning of the coil on a phantom. When compared to an equivalent quadrature coil, the SNR of the phased array coil was 35% higher than the quad coil when positioned in the center regions of the smaller elements, and 15% higher when positioned in the very center of the array [12]. The reduced gain at the center of the array occurred at points where elements overlapped and spins are in phase, causing the overall performance to tend toward that of a single large loop having the same perimeter as two array loops. Nevertheless, significant gains can be obtained using phased array coils at high fields. The increased frequencies of higher fields expose significant challenges in eliminating coil couplings and in the construction of the low impedance preamplifiers. The extension of these technologies to even higher frequencies represents a continuing and significant challenge.

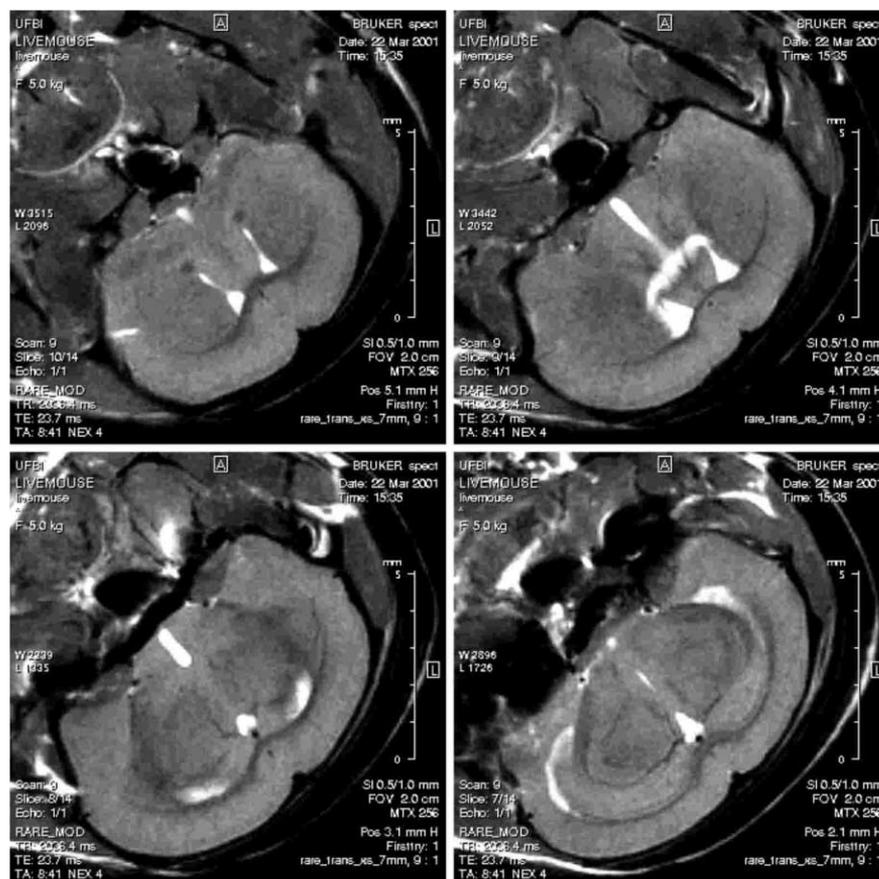


Fig. 4. Spin echo images of a live mouse at 750 MHz using the commercial birdcage coil. Resolution of  $0.78 \times 0.78 \times 0.5$  mm obtained in 8 h 41 min.

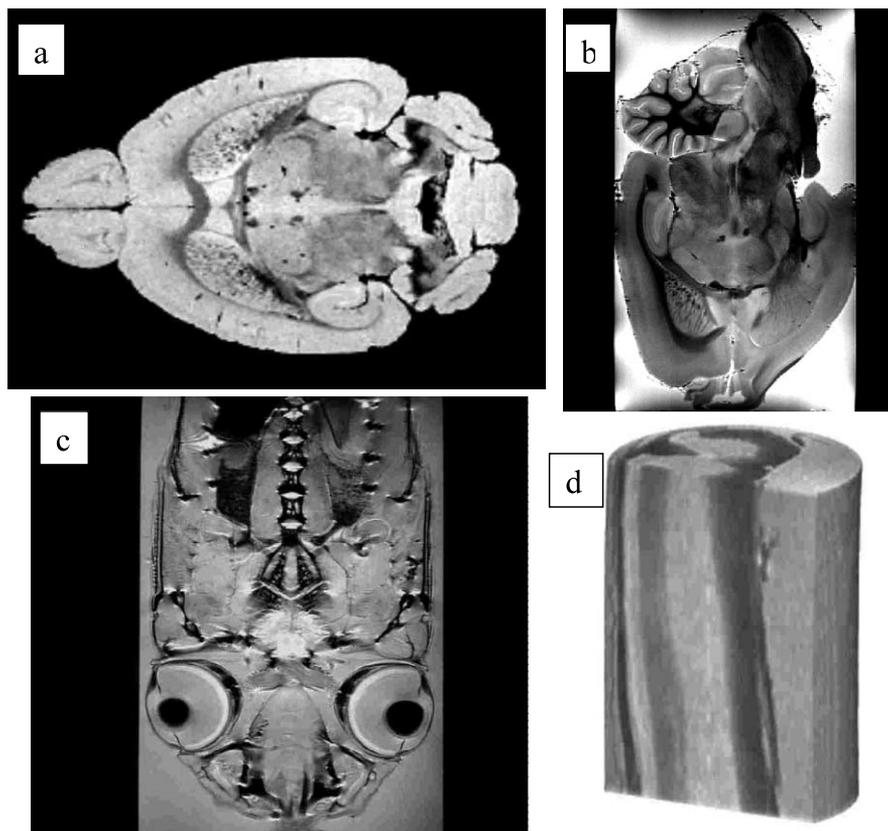


Fig. 5. Example images at 750 MHz of fixed tissue: (a) slice from a 3D mouse brain gradient echo image data set,  $47 \mu\text{m}^3$  in 2.7 h; (b) spin echo image of a rat brain, TR/TE = 5000/17 ms,  $60 \times 60 \times 200 \mu\text{m}$ , 11 h 22 min; (c) spin echo image of a small fish, TR/TE = 5000/15 ms,  $60 \times 30 \times 250 \mu\text{m}$ , 22 h 47 min; (d) 3D image data set of a section of rat spinal cord, TR/TE = 2500/16, 1 h 25 min,  $39 \times 39 \times 120 \mu\text{m}$  per voxel.

### 3.3. Microimaging and microcoils

The 750 MHz instrument also is intended to extend our microimaging studies, which range from samples as small as single cells up to adult mice. Using the commercial birdcage coils supplied with the instrument, our first images and localized spectroscopy of live mice have been obtained (Fig. 4). A full evaluation of the SNR of this instrument compared to lower fields is underway. The instrument has been used to image a variety of other samples, again without any significant difficulties (compared to imaging on lower field instruments). As examples, Fig. 5 shows example slices from 3D images data sets obtained of fixed mouse brains. The long-term goal of these mouse brain studies is to use the images to generate a probabilistic mouse brain atlas. The images show excellent contrast in these gradient-echo ( $T_2^*$ ) weighted images and have better SNR than we have achieved previously on similar samples at lower fields. However, a direct comparison is difficult when different systems, vendors, coils and techniques are employed. Consequently, a careful evaluation of the SNR is underway. Fig. 5 also shows other example image data sets on a rat brain, small fish and spinal cord.

Several of our studies depend critically on the SNR. We are limited in what we can present in this short paper, but indicate that the SNR is essential for accurate fiber track mapping of white matter in rat brains, which is being pursued via diffusion tensor imaging, and for the evaluation of the multicompartmental origins of diffusion signals in tissues. The latter is being pursued by diffusion studies on isolated brain slices [13,14], excised spinal cords [15] and a new red blood cell ghost model [16] on the 750 MHz instrument. SNR is also a paramount issue in our continuing microimaging and spectroscopy studies of isolated single neurons [17,18]. Even though these studies were performed to date at fields of up to 600 MHz, SNR improvements provided by even higher fields increase our ability to explore multicompartmental diffusion within the single cell [16] and to localize subcellular regions for spectroscopic examination [17] under a variety of perturbations. So far, bench tests and preliminary MR studies demonstrate no significant problems in constructing RF microcoils (solenoids having diameters of less than 1 mm) at 750 MHz. Thus, we look forward to improved SNR for single cell studies on this new instrument.

#### 4. Summary

In summary, as the drive towards higher magnetic fields continues, the development of the supporting technology must keep pace. We have shown that it is possible to construct the necessary additional hardware so that the advantages offered by the higher magnetic field can be fully realized. Both high-frequency volume and phased array coils appear feasible to construct and effective. Preliminary data using these coils are of high quality. At this point, it appears that other investigators would be justified in the pursuit of even higher field strengths, balanced by the additional costs that must be borne.

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