MR Imaging Using a Parallel Process Acquisition Concept: An Array of Super Conducting Coils for Space and Clinical Applications

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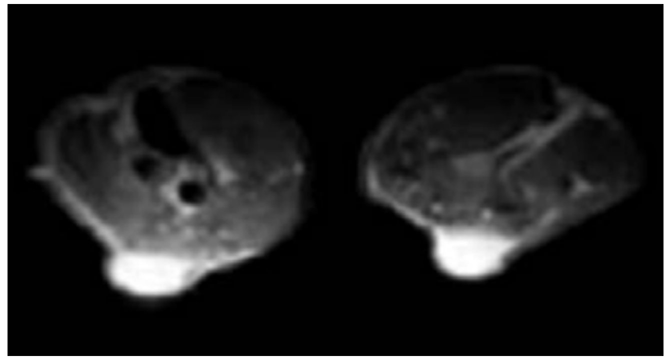
MRI IS A WIDELY USED DIAGNOSTIC TOOL THAT provides the unsurpassed ability to image soft tissue. In MRI, the subject is placed in a dc magnetic field and a sequence of field gradient and rf excitation pulses is applied. The relaxing nuclei (usually protons) produce weak decaying rf signals detected by an rf receiver probe.[1] Such signals are weak due to the small difference in energy levels population of parallel and anti-parallel spins (~6 parts per million at room temperature and at 1.5 T). In both research and clinical MRI, there is a need for high resolution and/or fast scan imaging; however, the signal-to-noise ratio (SNR) is the main limitation for realizing these needs. Therefore, the SNR is the most important parameter of MRI systems.

The overwhelming majority of MRI systems installed in diagnostic imaging facilities are costly and heavy mid-field (0.5-1.0 T) or high-field (1.5-3.0 T) systems. Fewer than one percent of MRI scanners in the U.S. are low-field systems, which operate below 0.15 T. The basic motivation behind using higher magnetic field strengths is that, in thermal equilibrium, the population difference between the spins in the upper and lower energy states is proportional to the strength of the dc magnetic field. In addition, each nucleus resonates at the frequency, which is proportional to the magnetic field, so that the amount of energy radiated at each transition increases accordingly. As a result, the MRI signal increases with the square of the magnetic field. The signal-to-noise ratio (SNR) will increase only linearly with the magnetic field, because the noise associated with eddy currents in the body of the subject increases linearly with the magnetic field. Noise in the system, in general, is created by conductive losses in the probe and in the body.

There are two regimes of such conductive losses in MRI systems.[3] In the first situation, the dominant loss is in the body, so that the SNR is body loss dependent; in the second instance, loss occurs when the SNR is primarily coil loss dependent. In the body-dominated regime, there is little advantage in the reduction of ohmic coil losses. However, when the coil loss is the governing source of noise, which is the case for low-field MRI (<0.35 T) or the case of small coils at higher fields, it has long been recognized that cooling the probe reduces this noise contribution and therefore can significantly increase the SNR of the measurement.[3]

MRI Coils—Dr. Jaroslaw (Jarek) Wosik, Research Associate Professor at the Texas Center for Superconductivity and Advanced Materials, focuses his research on medical applications. Funding on cryogenic MRI coils is sought to reduce the size of equipment for space flight and other portable applications.



IMAGING—The cryogenically cooled array (each coil is 1" in diameter) was used to acquire axial slices of subcutaneous tumors present in a murine model of colon cancer. Two animals were imaged simultaneously. One animal was under each element of the array, and both images were acquired simultaneously. The images were used to calculate tumor cross-section and observe tumor heterogeneity.

The cost and weight of MRI systems are points of great interest. A low-field MRI system, which can use either an electromagnet or a permanent magnet, is considerably lower than high-field units. However, the low SNR of low-field MRI systems has hampered their acceptance in radiology. UH researchers proposed to overcome this limitation and dramatically improve SNR with the use of high temperature superconductors (HTS) as a radio-frequency (rf) single receiver coil or as elements of receiver array of coils. HTS thin films are very attractive for use as surface receiver coils because at 77 K they exhibit an extremely low surface resistance R_s, several orders of magnitude lower than that of usually used copper.[4]

Development of low-field, low-weight MRI systems will provide support for JSC's primary tasks of human exploration, bioastronautics, and human operations in space. A small MRI system has been identified by the Science Working Group of the Human Research Facility on ISS as one of the most desired new research tools to be developed for documenting changes in muscle volume and assessing the effectiveness during long duration flights of in-flight muscle atrophy countermeasures. Establishing efficient musculoskeletal countermeasures was recognized as one of the critical path problems that must be solved in order to accomplish prolonged microgravity exposures required for a potential future Mars expedition. The development of a low volume, low mass, low energy imaging device, such as the MRI unit proposed here, will coincide with several sections of the NASA Critical Path Roadmap.[5]

An important step toward actualizing this proposal has been achieved by UH researchers who have designed and fabricated low-field MRI receiver coil (Fig.1). Such HTS rf probes provide a tenfold gain of signal-to-noise ratio (SNR) over standard copper coils. In Fig. 1, a low-field superconducting MRI receiver coil designed and fabricated by Dr. Wosik's group is shown.

In this ongoing project, UH researchers are working to constantly increase the SNR of the low-field system through the use of HTS for rf coils and array of coils fabrication. In order to further enhance the SNR and also simplify the scanner architecture future work, we will explore the use prepolarized MRI (PMRI) concept.[6] PMRI is a new MRI architecture based on field-cycled electromagnets suitable for low cost extremity imaging.

An important development in MRI technology has been the increasing success of MRI "phased" arrays as a method of increasing SNR and acquiring time reduction.[7,8] Most likely, future MRI scanners will use primarily arrays of surface coils, rather than volume and single surface coils.

In the frame of this project, we have developed a novel HTS-based design of a planar multi-layered structure (superconductor/dielectric/normal metal), which, in addition to HTS resonators, also includes built-in capacitors for tuning, matching and decoupling.[9,10] A double-sided structure was used to provide distributed capacitance for each coil resonator, in order to minimize stray electric fields that cause dielectric losses in the sample. Films were patterned into two split quasi-ring shapes (23-mm outer diameter, 17-mm inner diameter, and 15-mm shorter opening dimension). This concept allows for built-in decoupling capacitors between the coils in order to cancel mutual inductance.

Each coil was designed to provide a gain of at least 6 dB for this array over an identical room temperature copper array. The performance of the design was rf tested using different loss phantoms. The array was integrated with a custom-made G-10 liquid nitrogen cryostat (Tristan Technologies) designed for imaging of small animals.

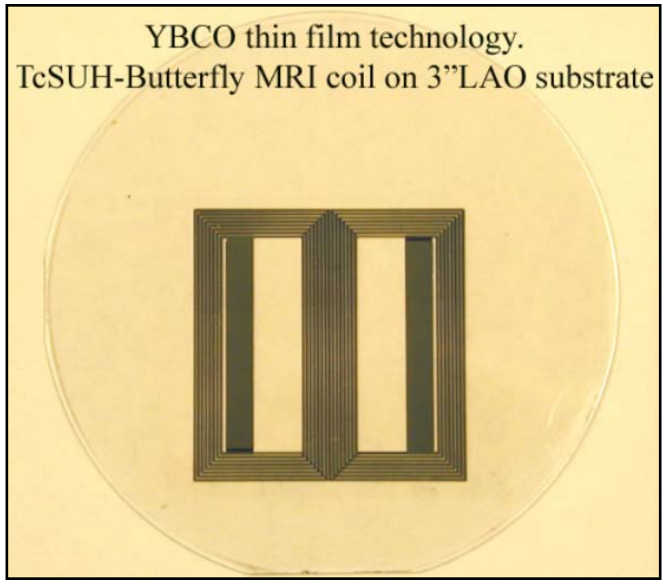


Figure 1. High-temperature superconductor (HTS) rf probes provide a tenfold gain of signal-to-noise ratio (SNR) over standard copper coils. The low-field MRI receiver coil designed and fabricated by the TCSAM group is shown.

Phantom loaded and unloaded quality factor measurements of the room temperature, 77 K copper array, and HTS array confirmed the expected SNR gain. The UH designed array of receiver coils also allowed the reduction by half of imaging time of two animals (see Fig. 2). Tests with animals were performed in the Department of Imaging Physics, The University of Texas M. D. Anderson Cancer Center in Houston, with the collaboration of Dr. John Hazle and Dr. Jim Bankson. Further tests of the array are now under way.

In Fig. 3, cancellation of the coils' mutual inductance is demonstrated by showing that each phantom image is acquired by one coil at a time.

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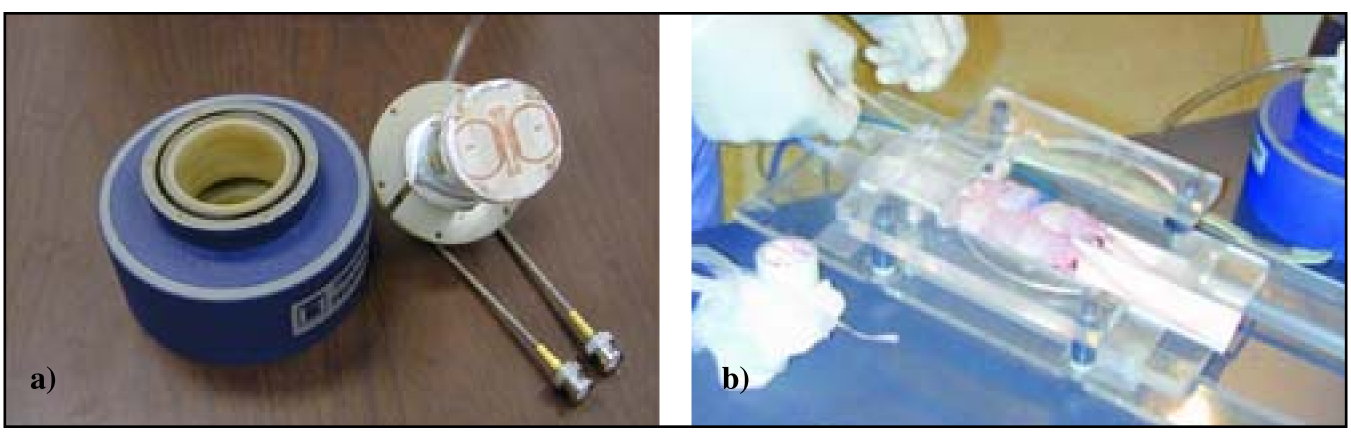

Figure 2. (a) The plastic, 77 Kelvin (liquid nitrogen) cryostat and the cryostat insert are shown with the upper side of an array clearly seen; (b) two mice are prepared under anesthesia and placed side by side for simultaneous MR imaging.

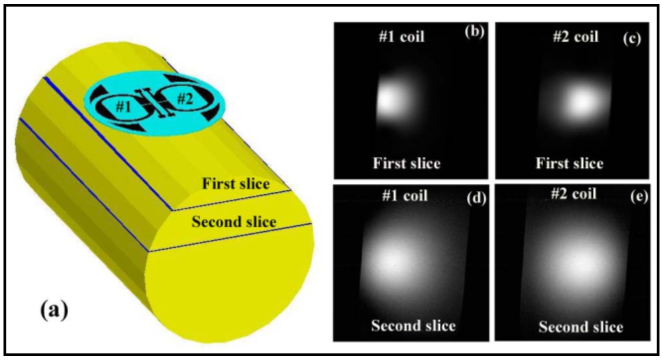

Figure 3. (a) The configuration of the array and phantom for the coil decoupling test is shown; (b) and (c) illustrate the first slice 4.7 Tesla images of the number one and number two coils, respectively; (d) and (e) illustrate second slice images of the number one and number two coils, respectively. Mutual decoupling of both coils can be clearly seen. Note that the #1 coil was placed toward the left side of the phantom.

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