EPR Imaging at a Few Megahertz Using SQUID Detectors

This frequency range is better for imaging of humans and large animals.

NASA's Jet Propulsion Laboratory, Pasadena, California

An apparatus being developed for electron paramagnetic resonance (EPR) imaging operates in the resonance-frequency range of about 1 to 2 MHz — well below the microwave frequencies used in conventional EPR. Until now, in order to obtain sufficient signal-to-noise radios (SNRs) in conventional EPR, it has been necessary to place both detectors and objects to be imaged inside resonant microwave cavities.

EPR imaging has much in common with magnetic resonance imaging (MRI), which is described briefly in the immediately preceding article. In EPR imaging as in MRI, one applies a magnetic pulse to make magnetic moments (in this case, of electrons) precess in an applied magnetic field having a known gradient. The magnetic moments precess at a resonance frequency proportional to the strength of the local magnetic field. One detects the decaying resonance-frequency magnetic-field component associated with the precession. Position is encoded by use of the known relationship between the resonance frequency and the position dependence of the magnetic field.

EPR imaging has recently been recognized as an important tool for non-invasive, *in vivo* imaging of free radicals and reduction/oxidization metabolism. However, for *in vivo* EPR imaging of humans and large animals, the conventional approach is not suitable because (1) it is difficult to design and construct resonant cavities large enough and having the required shapes; (2) motion, including respiration and heartbeat, can alter the resonance frequency; and (3) most microwave energy is absorbed in the first few centimeters of tissue depth, thereby potentially endangering the subject and making it impossible to obtain adequate signal strength for imaging at greater depth. To obtain greater penetration depth, prevent injury to the subject, and avoid the difficulties associated with resonant cavities, it is necessary to use lower resonance frequencies. An additional advantage of using lower resonance frequencies is that one can use weaker applied mag-
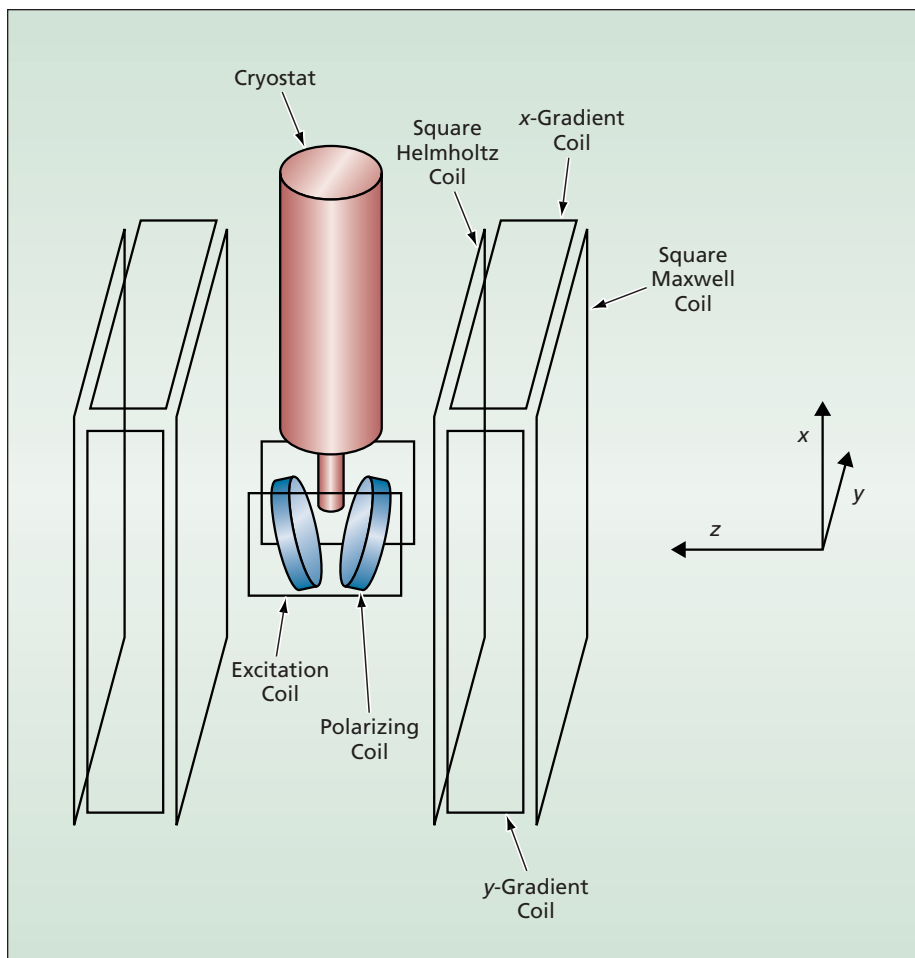
The **Electromagnet Coils and the Cryostat** of the present SQUID EPR imaging apparatus are depicted here in simplified form.

netic fields: For example, for a resonance frequency of 1.4 MHz, one needs a magnetic flux density of 0.5 Gauss — approximately the flux density of the natural magnetic field of the Earth.

In the present apparatus, a superconducting quantum interference device (SQUID) is used in detecting the EPR signal. Even without a cavity resonator, in the frequency range of about 1 to 2 MHz, it is possible to obtain a sufficient SNR by use of a simple pickup coil placed near the surface of the object to be imaged. The pickup coil is part of a flux transformer coupled to the SQUID. The signal is detected by either a commercial SQUID array amplifier device (which has a typical bandwidth of about 1 MHz) or a custom microwave SQUID (which has a typical bandwidth of about 10 MHz). To increase the SNR, the out-

put of the microwave SQUID can be fed to a cryogenic high-electron-mobility transistor amplifier.

The figure depicts the layout of the electromagnetic coils and the cryostat (which houses the SQUID circuitry) of the present apparatus. These coils are identical to those of a low-magnetic-field-strength SQUID MRI apparatus developed in the same laboratory. In one mode of operation, the polarizing coils are used to prepolarize the specimen by applying a magnetic field stronger than the nearly homogeneous magnetic field generated by the Helmholtz coils; the prepolarizing field is then rapidly turned off immediately before starting to record EPR imaging data. The polarizing coils are positioned and oriented so as not to induce measurable flux in the pickup coil. It is important to note

Reducing Field Distortion in Magnetic Resonance Imaging
An unconventional magnetic-field configuration would be used in SQUID MRI.

*NASA's Jet Propulsion Laboratory, Pasadena, California*

A concept for a magnetic resonance imaging (MRI) system that would utilize a relatively weak magnetic field provides for several design features that differ significantly from the corresponding features of conventional MRI systems. Notable among these features are a magnetic-field configuration that reduces (relative to the conventional configuration) distortion and blurring of the image, the use of a superconducting quantum interference device (SQUID) magnetometer as the detector, and an imaging procedure suited for the unconventional field configuration and sensor.

In a typical application of MRI, a radio-frequency pulse is used to excite precession of the magnetic moments of protons in an applied magnetic field, and the decaying precession is detected for a short time following the pulse. The precession occurs at a resonance frequency proportional to the strengths of the magnetic field and the proton magnetic moment. The magnetic field is configured to vary with position in a known way; hence, by virtue of the aforesaid proportionality, the resonance frequency varies with position in a known way. In other words, position is encoded as resonance frequency.

MRI using magnetic fields weaker than those of conventional MRI offers several advantages, including cheaper and smaller equipment, greater compatibility with metallic objects, and higher image quality because of low susceptibility distortion and enhanced spin-lattice-relaxation-time contrast. SQUID MRI is being developed into a practical MRI method for applied magnetic flux densities of the order of only 100 μT.

In conventional MRI, it is required that all of the magnetic moments precess around the same axis defined by an applied static, homogeneous magnetic field. One of the problems in developing low-magnetic-field-strength MRI arises from the relationship among image resolution, image-acquisition time, and the gradient of the magnetic field. In general, the resolution (that is, the reciprocal of the smallest size of an
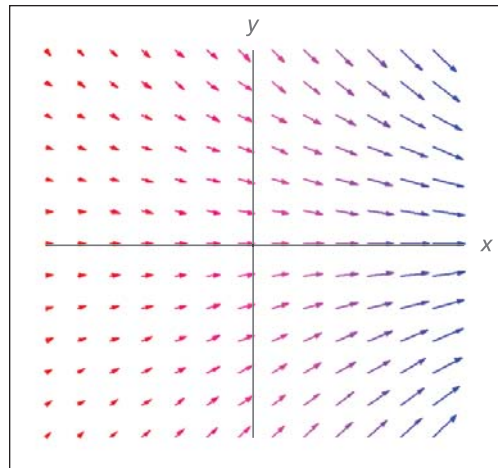
Field Configuration is shown in *x,y* plane.

image feature that can be resolved) is proportional to the strength of the magnetic-field gradient and the image-acquisition time. Because of fundamental properties of a magnetic field, an image-position-encoding gradient necessarily entails undesired concomitant field components proportional to the image-position-encoding gradient. These components are undesired because they cause distortion or blurring of the image. The distortion can be reduced by weakening the gradient so that the total field variation in the magnetic field is a small fraction of the static homogeneous field. However, in weakening the gradient, one must either accept lower resolution or else increase image-acquisition time to retain resolution.

According to the present concept, it should be possible to reduce distortion to a very low (in principle, negligible) level while using a low-strength magnetic field, without sacrificing resolution and without need for excessively long image-acquisition time. There would be no applied static, homogeneous field because in MRI, it is not required that all the magnetic moments precess about the same axis. The encoding magnetic field, **B**, would be configured to lie in the *x,y* plane of *x,y,z* Cartesian coordinate system: the field would be given by

$$\mathbf{B} = B_0 e^{x/a}[\hat{x}\cos(\mathbf{y}/a) - \hat{y}\sin(\mathbf{y}/a)]$$

where $\hat{x}$ and $\hat{y}$ are unit vectors along the *x* and *y* axes and B_0 and *a* are arbitrary constants to be chosen by design (see figure). Inasmuch as the strength of this field is given by $|\mathbf{B}| = B_0 e^{x/a}$, this field would encode position along the *x* axis as a function of resonance frequency. Although *x* would be a nonlinear (specifically, a logarithmic) function of resonance frequency, it would be possible, through suitable choice of the arbitrary constants, to reduce the deviation from linearity and the concomitant spatial variation of resolution over the image region to tolerable levels.

A close approximation of the magnetic field described above could be generated by electric currents flowing axially on a *z*-oriented circular cylinder having a length much greater than its radius. The current density (*K*) needed to generate the magnetic field is a known continuous function of the azimuthal angle (θ). In practice, the required continuous current density *K*(θ) would be approximated by use of multiple discrete wires positioned at suitable angular intervals and driven by suitable currents. To encode position along an axis at an angle α with respect to the *x* axis in the *x,y* plane so that back projection reconstruction could be used, one would shift the cur-