INTERNATIONAL ATOMIC ENERGY AGENCY
UNITED NATIONS EDUCATIONAL, SCIENTIFIC AND CULTURAL ORGANIZATION
INTERNATIONAL CENTRE FOR THEORETICAL PHYSICS
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H4-SMR 471/16

COLLEGE ON MEDICAL PHYSICS

10 - 28 SEPTEMBER 1990

MRI INSTRUMENTATION - PART I

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I. Introduction

Magnetic resonance imaging (MRI) has been proven to be a valuable diagnostic imaging modality in assessing pathology of the brain, spinal cord, musculoskeletal system, abdomen, heart, and great vessels. Compared to virtually all other medical imaging modalities, MRI is unquestionably the most technologically complex. It requires special purpose equipment for signal acquisition as well as powerful high speed computers for image reconstruction. The purpose of this lecture is to describe some basic concepts and characteristics of the instrumentation used for magnetic resonance image formation. As a reference we use the schematic diagram of a generic MRI system which is shown in Fig. 1. For purposes of this discussion we somewhat arbitrarily divide the entire MRI system into three major parts: (i) the magnet, (ii) instrumentation used for creation of the MRI signal, and (iii) instrumentation used for converting the MRI signal into an actual image.

II. The Magnet

The magnetic field of the MRI system is required to create the differences in population of energy levels of nuclear spins. For a total population of N spins the net magnetization M_o which is created is given by:

$$M_o = N \mu B_o^2 / k T \qquad [1]$$

where B_o in the field strength, μ the magnetic moment of the nucleus of interest, k is Boltzmann's constant, and T is absolute temperature. Eq. 1 shows that the net signal increases in proportion to the square of the applied field strength. This suggests that the higher the field strength then the more favorable are the conditions for imaging. This is offset somewhat by a modest increase in noise with increasing field. Ultimately, inadequate penetration of the radiofrequency (RF) fields necessary for NMR signal measurement is expected to limit the usable field strength for human imaging to about 4.0 Tesla or less. Starting from high fields (4.0 T) and working downward, the magnets used for MRI are superconducting in the high field range (~1.0 to 4.0 Tesla), resistive in low to mid field (up to 0.4 T), and permanent magnets are possible for fields less than approximately 0.4 T.

Just as for standard NMR spectrometers, a major requirement of the magnet used for imaging is stability. One can place a specification on stability by requiring that the shift ΔB in field, and hence the shift in Larmor frequency, be small (say less than 5%) compared to the bandwidth across one pixel:

$$(\gamma/2\pi)\,\Delta B \le 0.05\,\Delta f / N \qquad [2]$$

Here $\gamma/2\pi$ is the gyromagnetic ratio, Δf the bandwidth of the received signal, and N the number of pixels across one direction of the image. Solving for the fractional homogeneity $\Delta B/B_o$:

$$\Delta B/B_o \le 0.05\,\Delta f / N\,(\gamma/2\pi)\,B_o \qquad [3]$$

and inserting nominal values yields a specification of 0.1 ppm.

In addition to stability other major specifications pertain to homogeneity, field of view, and shimming capability. The homogeneity is a measure of the ability to resolve two peaks in the

NMR spectrum. For high field systems of 1.5 Tesla the homogeneity spec for imaging over a 20 cm diameter volume is typically 0.5 ppm or less. For in vivo spectroscopy this ideally decreases by an additional factor of three or more. The largest fields of view (FOV) encountered in MRI are in imaging of the thorax and abdomen where 50 cm fields are possible. It is desirable in this case that homogeneity still be near that for the 20 cm FOV case. Finally, shimming refers to the ability to superimpose via additional coils small magnetic fields which are corrections to the inhomogeneity of the intrinsic field of the magnet. Typically these are designed as higher order polynomial functions, and it is common to have as many as 11 such terms.

III. MR Signal Creation

This major section of the instrumentation is that which enables spatial encoding and nutation of the magnetization vector.

Gradients. The purpose of the magnetic gradients is to provide spatial encoding of the magnetization. In accordance with the Larmor equation, the application of a gradient along, say, the x direction causes the resonant frequency f to vary along x:

$$f(x) = (\gamma/2\pi)\,(B_o + G_x\, x) \qquad [4]$$

Similarly, gradients are necessary for the y and z directions as well. It should be stressed that these gradients do not cause a magnetic field to be generated in the x or y directions, but rather the z-directed magnetic field can now vary in the x, y, or z directions depending upon which gradient is applied.

Major specifications pertaining to gradients are linearity, amplitude, rise time, and the generation of eddy currents. Linearity is largely dictated by the geometry of the gradient coils themselves. The z gradient set is composed of a pair of helically wound coils of opposing orientation. The x set is comprised of Golay coils placed around a cylindrical plexiglas form. the y set is identical to the x set except it is rotated 90° about the z-axis. For gradients used with whole body MRI units the typical maximum amplitude available presently is 1 Gauss/cm. This is adequate for many conventional spin-echo imaging sequences. With resonant circuit technology this can be larger by about a factor of four. Rise times for gradients are dictated by the inductance of the coil and the applied voltage. A common value is 500 μsec from 0 to 1 Gauss/cm. Smaller times are desirable for high speed scanning sequences. Finally, the application of a current through a gradient coil can induce eddy currents in the cryostat which in turn cause magnetic fields which oppose the intended applied field. This perturbs the net gradient field applied and can result in image artifact. This can be counteracted by using self-shielded gradient coils in which an outer coil set opposes at the cryostat the magnetic field created by the inner coil. This method of "self-shielded" coils works effectively.

Radiofrequency (RF) Amplifier/Coil. The purpose of the RF section during acquisition is to apply radiation at the Larmor frequency to the sample under study:

$$f = (\gamma/2\pi)\, B_O \qquad [5]$$

For example, for imaging protons at a field of 1.5 Tesla, the resonance frequency is 63.5 MHz. Major requirements of the RF system include tuning capability, and control of the modulation of the RF. Tuning includes both adjustment of the RF frequency to account for slight patient-to-patient variations as well as of the RF amplitude to allow for different nutation angles or sample sizes. It is necessary to modulate the RF waveform in order to generate so-called "selective" RF pulses which when applied simultaneously with a magnetic gradient cause only a selected region in the object to undergo nutation.

Master Sequencer. An MRI pulse sequence involves the repetitive application of a set of waveforms to each of several devices. These include the three gradient amplifiers, the real and imaginary channels of the RF amplifier, and possibly additional channels for control. This is typically handled by a master sequencer or some equivalent set of electronics. This sequencer causes all waveforms to be synchronized to each other and insures that all data is created in a very carefully controlled, reproducible fashion. Exact designs vary from one manufacturer to another. However, the temporal resolution required is of the order of 1 μsec. The bit resolution or precision of the generated signals is typically 12 bits or higher.

IV. MR Image Formation

The purpose of this major set of subsystems is to provide for the measurement of the NMR signal, the storage of such signals, and the reconstruction of the MR image.

RF Receiver Coil/Preamp. The purpose of the RF receiver is to detect the transverse magnetization which is refocussed in the form of an echo. In many cases the same coil can be used for receive as is used for transmit. In other instances it is possible to use custom receiver coils which are often placed on the surface of the subject to be in close proximity to the volume of interest. The preamp amplifies the detected signal from the μV to the mV range for transmission to the subsequent receiver circuitry.

Demodulator/Digitizer. The received NMR signal is modulated at the Larmor frequency as determined from Eq. 5, 63.5 MHz for protons at 1.5 Tesla. In order to form an image it is first necessary to remove this carrier frequency. This is done using the standard technique of demodulation and results in a signal whose bandwidth is determined by the readout gradient. By simple modification of Eq. 4 one can show that Δf is given by:

$$\Delta f = (\gamma/2\pi)\,(G_x\ FOV_x) \qquad [6]$$

where FOV is the field-of-view along the x direction. Using typical values yields a bandwidth of ±15 kHz. This in turn requires digitization rates of 30 kHz or more for both the real and imaginary channels of the received signal.

Storage and Reconstruction. MR image data is commonly acquired via a series of 128 or more applications of one pulse sequence with a repetition time between applications in the range of 20 to 3000 msec. During each repetition one signal is measured, a typical signal being 8 msec long and digitized to 256 samples. Only after all signals have been acquired does the reconstruction process commence. This requires adequate storage for the acquired data. With the interleaved acquisition of data for up to 20 sections, the intermediate storage requirements can run up into the

10 MByte range.

Conversion of the MRI acquired data into an actual image involves the application of a two- or three-dimensional Fourier transform operation. This is performed with an array processor (AP). Modern APs can execute 256x256 image reconstruction algorithms in times under one second. The AP can also be vital in rapidly performing other steps in the image formation process, such as correction for system non-idealities.

V. Summary

The instrumentation required for modern MRI systems encompasses many aspects of physics and engineering, ranging from electromagnetics and RF design to the development of custom microprocessor circuits and microcoded software algorithms. It is only through the efforts of physicists and engineers that MRI systems have proven to be so useful as they are today.

Fig. 1 Schematic of MRI system.