Nuclear Magnetic Resonance Medical Imaging

Pages with reference to book, From 64 To 75
Syed Farooq Akber (Department of Radiologicnl Sciences, UCLA School of Medicine, Los Angeles, California 90024.)

Abstract

Nuclear magnetic resonance medical imaging is a new diagnostic tool available to radiologist for investigating the morphology and pathophysiology of the human body. In the presence of constant, stable, and uniform magnetic field and the application of radiofrequency radiation, will enable to produce anatomic cross sections. The physical principles of nuclear magnetic resonance medical imaging as well as the scanning procedures are described. The clinical applications of nuclear magnetic resonance medical imaging are also discussed (JPMA 34 : 64, 1984).
Key words : NMR, NMR Imaging.

Introduction

Nuclear magnetic resonance •(NMR) principles and theory were established 35 years ago by Purcell et at. (1946) and Bloch et at. (1946), independently. The idea of NMR imaging was first proposed by Damadian (1971) in 1972. It was not until 1973, however, that Lauterbur (1973) conceived the ldea of using NMR principles to reconstruct two dimensional transverse images generated from four one dimensional, projection on to a phantom. The reconstruction approach used was similar to back projection technique currently employed in computed tomography (CT). NMR is a versatile imaging modality because it is non-ionizing, non-invasive, and without known biological risk. In addition, NMR provides images of high spatial resolution comparable to x-ray computed tomography (XCT) and allows tomographic imaging based on the chemistry and metabolism of the human body. NMR's success in visualizing the diseases affecting soft tissues, such as cancerous tumors; demonstrates vast potential to diagnose heart disease or stroke at an early stage by monitoring the blood flow to the heart or brain. Calcified tissues and air, on the other hand present no obstacle to satisfactory image production and soft tissue differential.

In the last couple of years NMR imaging has appeared on the medical imaging scene (Gore et al., 1981; Dixon and Ekstrend, 1982; Lerski, 1983; Gykett, 1982; Pykett et al, 1982; Pennock, 1982; Hinshan and Lent, 1983). It is most likely that NMR become a standard diagnostic technique and may supersede x-ray computed tomography. In this paper, the physical principles of NMR imaging and its clinical significance and potentials are discussed.

Physical Principles

NMR is a nuclear phenomena rather than an atomic one. The nucleus of an atom is positively charged and contains neutron and proton particles. The neutron is aneutral particle. The proton is positively charged and possesses angular momentum due to its spin. The spinning of a proton about an axis through its center corresponds to a current of positive electricity flowing in a circle around the axis of spin. Such a circular current should have a magnetic field associated with it and hence can be represented by a small dipolar magnetic moment. The strength and direction of the magnetic field can be determined by a vector called magnetic moment. Nuclei with odd numbers of neutron and proton (or both) exhibit magnetic properties and are identified by NMR spectroscopy. By contrast, nu.clei with an even nunibers of neutrons and protons (even-even nuclei) exhibit zero angular momentum and do not have magnetic properties and hence does not generate NMR signals.

The hydrogen nucleus consist of a proton only; this proton has a half spin. Hydrogen nuclei in a zero magnetic field orientated randomly and align itself parallel or antiparaliel when magnetic field H_o is

applied as shown in Fig. 1.

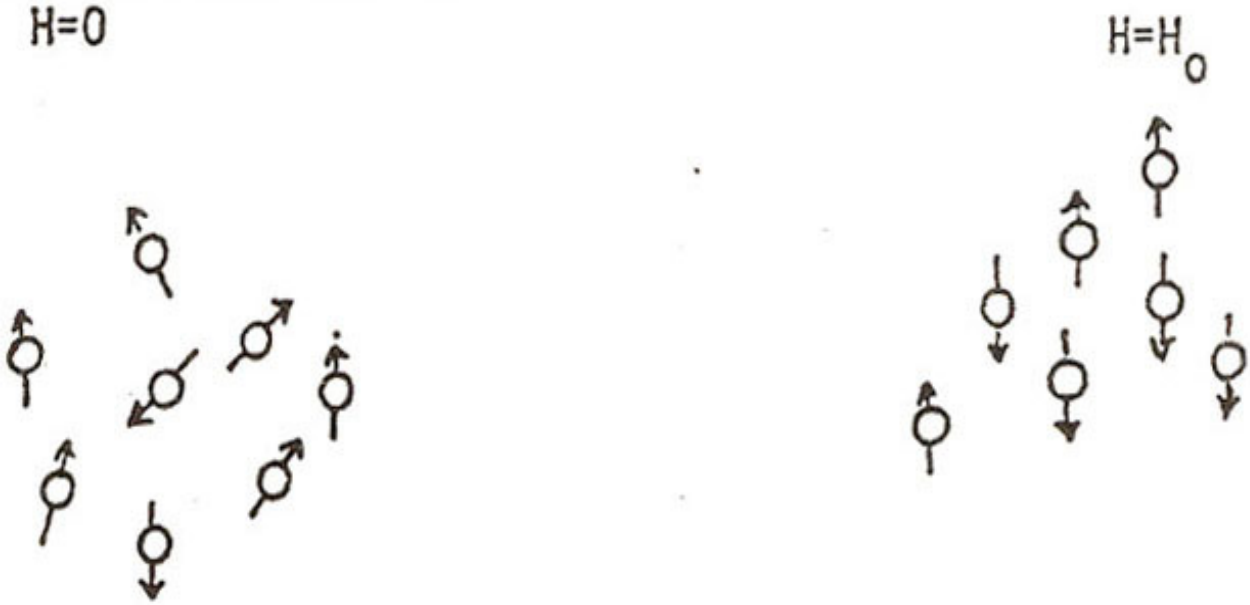

Fig. 1. Orientation of proton spins in zero magnetic field and in an applied field H_o.

The opposite direction with respect to H_o. When the RF of precession of H_1 is equal to Larmor frequency, then the resonance will occur and this phenomena is called nuclear magnetic resonance. The stimulating energy H_1 to flip the net magnetization through an angle 0 with respect to is commonly known as 0 pulses. Generally, the applied pulses are either 90^o or 180^o and is referred to as 90^o pulse or 180^o pulse. The pulse separation of protons into two groups, also produces a difference in energy between them. Protons population difference with spin "up" (parallel) and spin "down" (antiparallel) is however, very small; with a slight excess of popula. tion in the lower energy state spin "up"
A hydrogen nucleus as mentioned earlier has angular momentum. As a consequence. spin axis proton will align itself in the presence of static magnetic field H_o but because of thermal agitation do so incompletely and precess at an angle 0 about H_o. This precession. generally known as Larmor precession (Fig. 2)

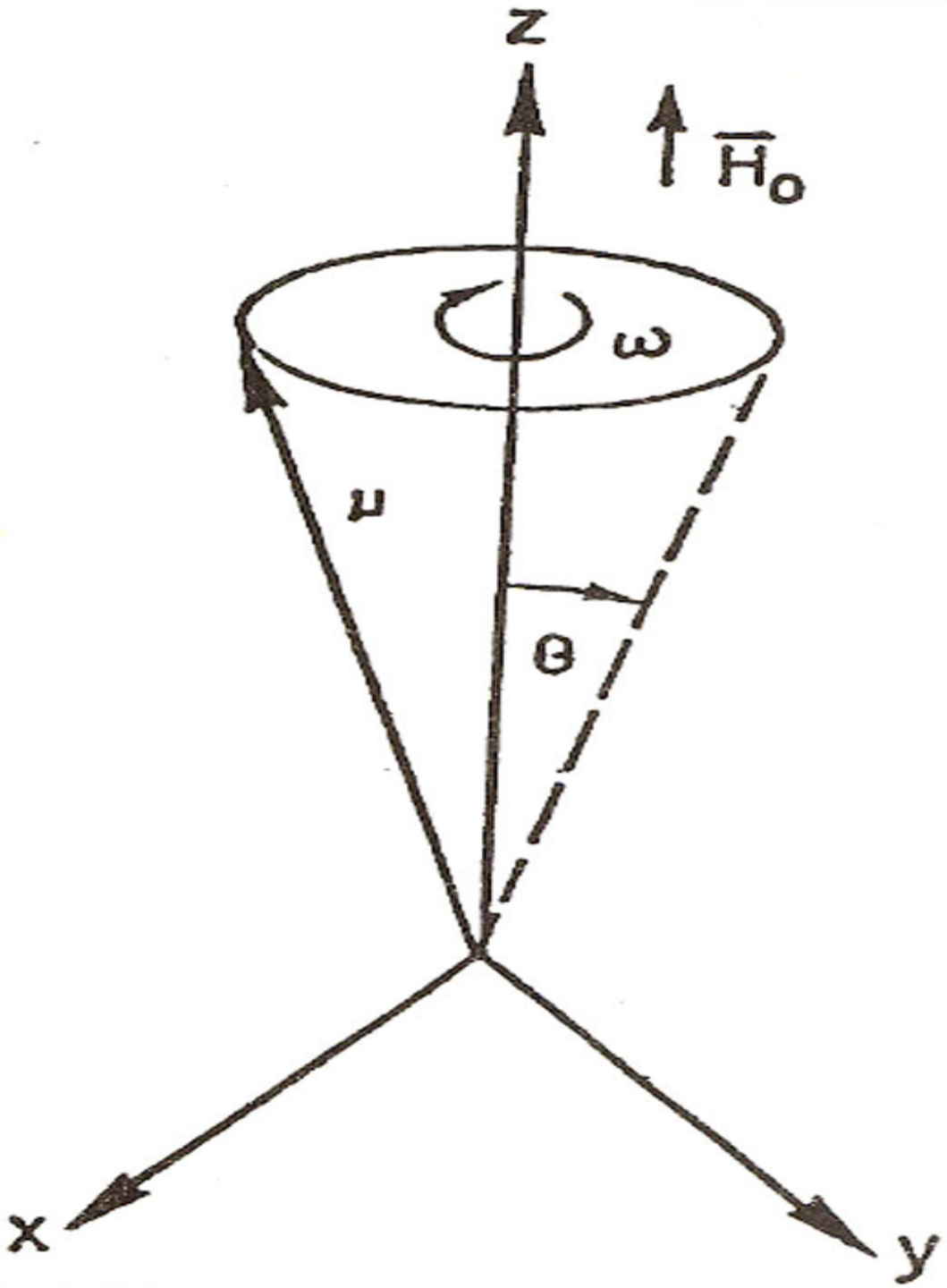

Fig. 2. Spinning of a proton with a magnetic moment, μ, At an angular velocity, ωat an angle θ, with the external Field $\bar{H}_o$z.

is given by a funda mental equation w =Y H_o, where V is known as gyromagnetic ratio, and 0 is known as Larmor frequency. This Larmor precession is identical to the precession of a gyroscope about a gravitational field. The precession occurs as a result of the interaction between the torque generated by the rotational motion of the nucleus and the interaction of the magnetic field on the nuclear magnetic

moment.
The application of a RF excitation energy in the form of a small magnetic field H_1 on a precessing proton will flip the magnetic moment in J.P.M.A. March, 1984 duration for excitation of spinning proton is 50 to 5000 microsecond. Nuclei will absorbed energy from the stimulating field and they will precess about H_o at a larger angle of 0, depending on the strength of RF field H_1. The Larznor condition can be monitored by measuring the absorption of energy from the stimulating field. When the RF field is ceased, the nuclei will tend to return to its original position, i.e. to precess along H_o. In doing so, the field emits a RF wave of the same energy as that absorbed. These waves can be externally detected by RF coils enclosing the patient, generates the NMR signal that can be reconstructed three dimensionally to form an image.
The resonance phenomena under a fixed and uniform magnetic field is artificially dependent upon the RF and must be identical to the Larmor frequency. Variations in either the strength of the magnetic field or the applied field will preclude resonance. Because nuclei have different values of magnetic moment and spin, resonance occurs at varying frequencies within fixed fields. This principle is used to map the nucides such as hydrogen, carbon 13, fluorine 19, sodium 23, and phosphorous 31. Hydrogen is generally used for imaging because of its abundance in the human body. Hydrogen nuclei's gyromagnetic ratio is 42.577 MHZ/Tesla and the resonance frequency is 3.4 MHZ in a 0.1 Tesla field. Mapping other isotopes inside the body is a difficult task, since the concentration of these nuclides are very)bw and do not generate adequate signals for high resolution images. However, great interest in mapping phosphorous 31 has been generated and great strides have been made in this respect (Evanochiko et al., 1983).

NMR Measuring parameters

Five characteristics of the human body can be measured by NMR.
1. Coicentration of nuclei, such as hydrogen, carbon 13, fluorine 19, sodium 23, and phosphorous 31.
2. When the RF pulse is ceased, the magnetization vector tends -to return to its original position parallel to the static magnetic field Ho and the emitted signal decreases in strength with time. This phenomena is characterized by two relaxation times, T_1 and T_2.

T_1. Spin-lattice relaxation implies the transfer of energy to the aggregate of atoms in the form of thermal energy. The spin-lattice relaxation time (T_1) is the time required for the bulk magnetization state of the material to return to equilibrium after RF stimulation. It is dependent on the viscosity, temperature, and concentration of the material.
3. T_3. After an RF pulse has tipped the nuclear magnetization vectors towards the transverse plane XY, the components of this vector phase coherence is lost and some nuclei precess at different rates than the others due to fluctuations of the magnetic field strength. RF waves emitted from individual nuclei are out of phase and they cancel out each other, so the sum total of nuclear magnetization vector in the transverse (XY) plane decay to zero. The total time characterizing this dephasing or decay is called T_2*. This time is further reduced in NMR imaging system by using magnetic field gradient.

T_2 which is the intrinsic component of the signal decay of the sample, is a result of energy interaction between spinning nuclei, hence it is called spin-spin relaxation (Fig.3).

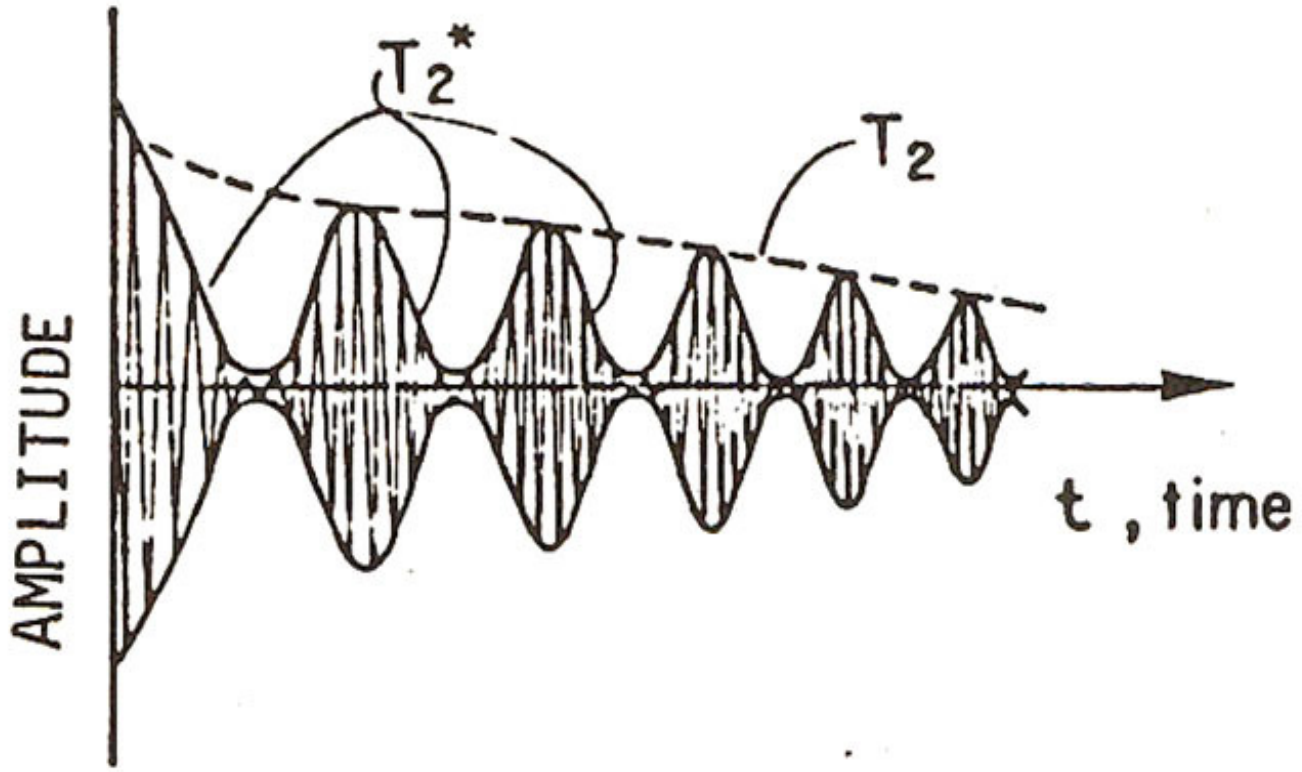

Fig.3. T_2 is the Decay value of the spin-spin relaxation Time and is the Decay time. T_2* is the Decay time for the individual Echo.

This time implies the transfer of energy directly to neigh. boring nuclei. This measurement is influenced by temperature and pressure changes.

Generally, T_1 is always greater than T_2. In solids, T_1 is longer than T_2, e.g., in water T_1 and T_2 are approximately equal (3 seconds). In ice, T_2 is in minutes and T_2 is in microseconds.

4. Chemical shift. Following an RF irradiation, nuclei re-emit this RF energy and the resulting signals can be detected for analysis. Variations in the molecular environment cause the resonance condition of the nuclei to relax, or give up their energy at different rates, a phenomenon called chemical shift.

5. The change in signal intensity due to movement of magnetized nuclei through the resonance region depends on volume information.flow

NMR Imaging System

NMR imaging system block diagram is given in Fig 4,

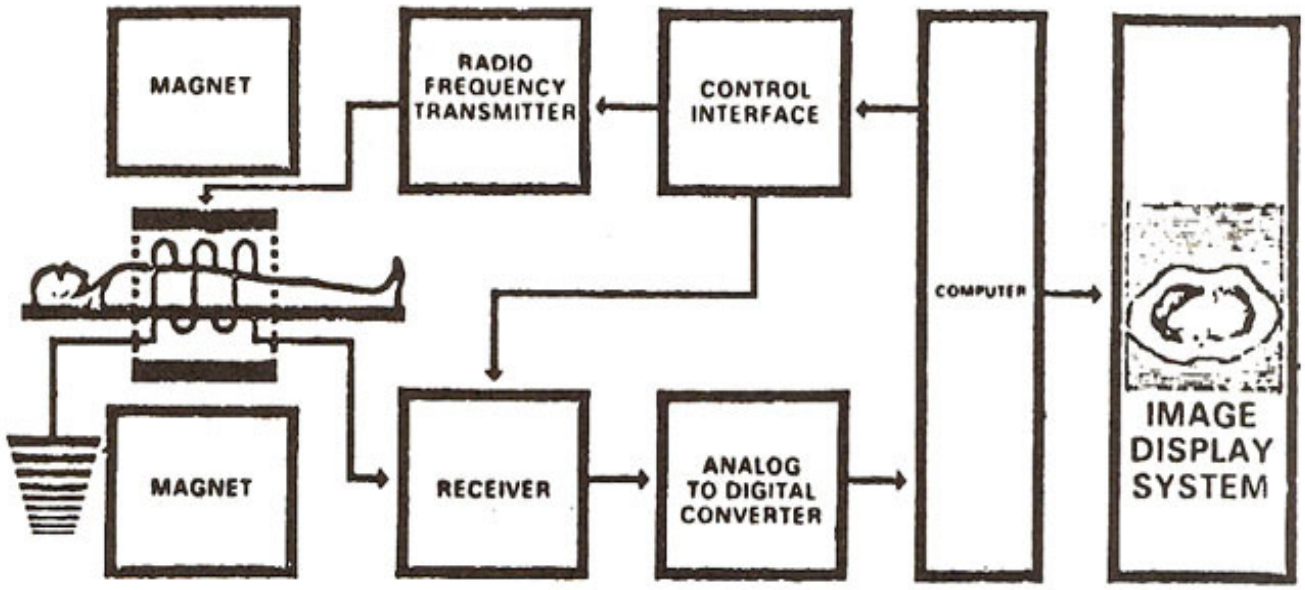

Fig. 4. NMR Imaging System.

and the basic component of the system (Partain et al., 1983) is given in

Table 1

General Features of NMR Imaging System.

1.	MAGNET SYSTEM
	Static field generation coils
	DC Power supply
	Cooling system (resistive systems only)
	Gas recovery (Refrigeration super conducting system only)
	Active and passive shimming mechanism
	Gradient coils —x,y,z sets
	RF coils
	Transmitter
	Receiver
	RF coil tank circuits(matching-networks)
	NMR preamplifier
	Magnet shielding and power lead filters
	Patient handling
2.	SPECTROMETER
	Stable RF source
	Transmitter (pulse forming circuitry)
	Receiver (amplification and demodulation)
3.	DATA ACQUISITION SYSTEM
4.	GRADIENT SYSTEM
	Wave form generation
	Power amplifiers
	Switching (e.g., for selection of planes)
5.	RF AMPLIFIER
6.	LOGIC
	Control of analog functions
	Pre-FT arithemetic
	Post-FT arithemetic
	Computer interfacing
7.	COMPUTER AND DISPLAY SYSTEM
	Basic CPU
	Peripherals
	Display System
	Operator input facilities

Table I and are briefly discussed below. In NMR medical imaging, the patient in a clinic lies on a moveable bench that is guided into a large circular gantry with axis of the gantry usually along the long axis of the patient. The gantry houses a massive magnet to produce a strong and stable static axial magnetic fields, coils to produce gradients in this field and an RF coil to transmit the stimulating pulses

and detect the NMR signal, which is reconstructed by computer to give cross sectional images.
1. MA GNET. The heart of any NMR system is its magnet. The magnet can be either resistive or superconducting, so long as it provides a strong, uniform and steady magnetic field.

Table - II Magnet Characteristics.

	Resistive	Superconductive
Field strength (Upper Limit)	0.2T max	2T max
Practical	0.1 - 0.15 T	0.35 (highest in current use)
Power dissipation	15 kW (0.1T) 60 kW (.2T)	Zero
AC	27 KVA 100 KVA (30)	? (low)
Cooling	closed water cycle	None
Homogeneity	1 in 10 easy	1 in 10 easy
Homogeneity DSV	45 cm max typ	50 cm easy – greater at no penalty
Stability (short term)	1 in 10 max 1 in 10 likely	1 in 10 possible
(long term)	2 hour warm up	say 10 ppm
Advantages	low cost case of use safety	homogeneity for DSV better short term
Disadvantages	High power dissipation	Cost

Table II gives the characteristic of both resistive and superconducting magnet (Holland, 1983).
2. RF TRANSMIITER AND RECEIVER. This component generates radiofrequency magnetic fields to the sample and detects signals.
3. GRADIENT SYSTEM The purpose of this component is to select a uniform slice thickness, which inturn generates signal from thin slice. This system produces time.varying magnetic field of controlled spatial non uniformity.
4. DETECTION SYSTEM Yields the output signal.
5. 1MA GING SYSTEM Includes computer that stores data and reconstructs and displays the images.
The nature of information abstracted in the NMR image depends on the sequence RF pulses used to perturb data before acquisition. The three basic pulse sequence used for clinical imaging
a. The repeated free induction decay sequence.
b. The inversion recovery sequence
c. The spin echo sequence.

Free Induction Decay

The repeated free induction decay (FID) sequence is shown in Fig. 5,

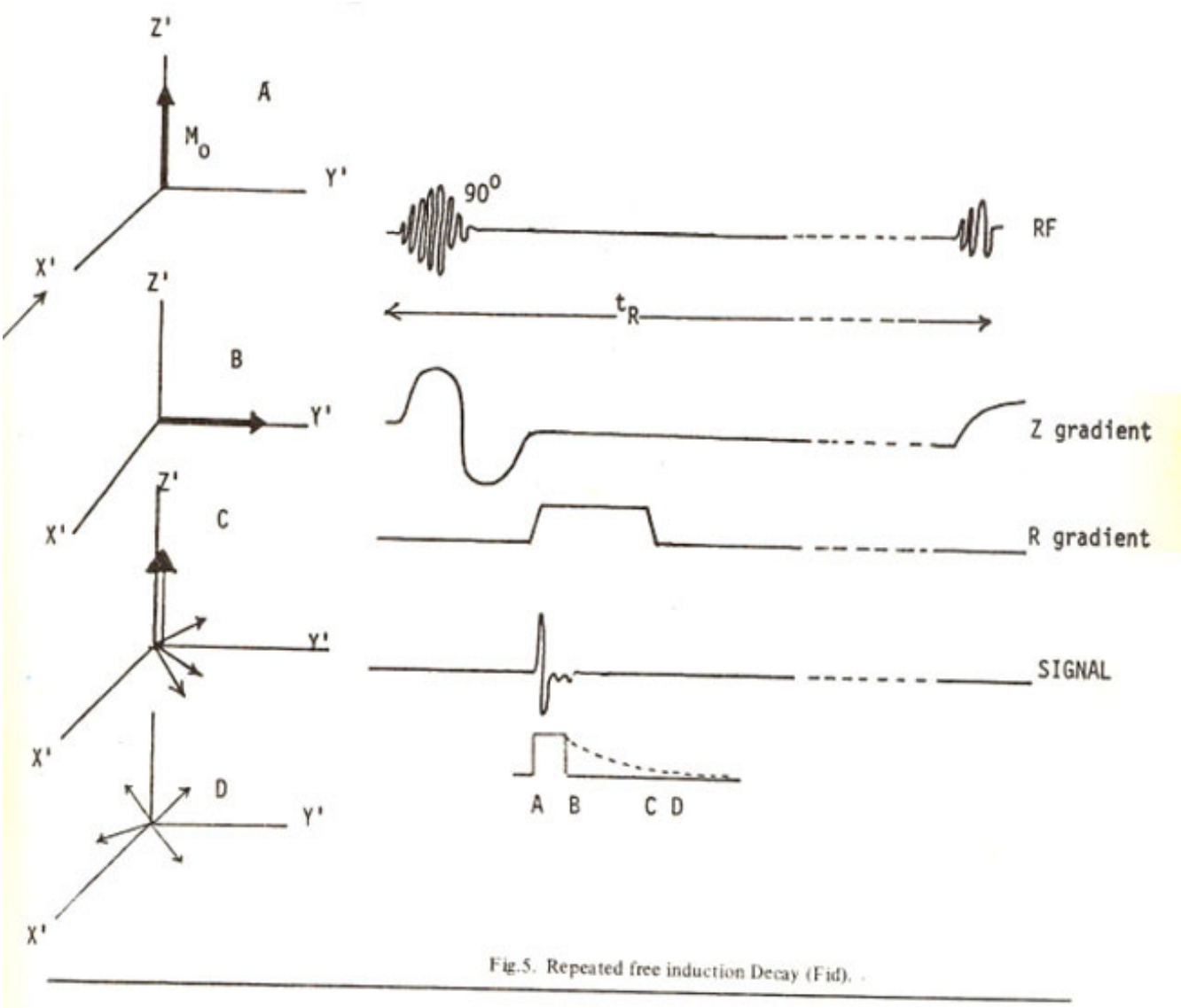

Fig.5. Repeated free induction Decay (Fid).

in which the magnetization M_o in a frame rotating with the properly precessing component of H_1.

An FID signal is produced using a 90_o RF pulse which is sampled and then another 90_o RF pulse is applied after a time interval of tR before the magnetization reaches its equilibrium value of Mo along the +Z axis. This inturn initializes M2 to zero after the 90^o RF pulse, recovered to a value

$M_z(t) = M_o (1\text{-}e\ tr/T_1)$

at time tR, jnat before the next 90_o pulse the transverse component value of M in the frame rotating after the 900 pulse rotates Mz on to the Y' axis is

$M_{y'} = M_o (I\ \text{-exp-}tR/T_1)$

To improve the data collection efficiency, the second 90_o RF pulse sequence should be applied instantaneously not allowing the complete recovery of the magnetization to its equilibrium position. In order to achieve an optimum efficiency, the delay time between two 90_o RF pulses should be tR roughly equal to T_1 (Hinshaw and Lent, 1983).

FID sequence produces a strong signal dependent on proton density represented by M_o. Image representation is the depiction of T_1 and image pixel value is proportional to proton density.

Inversion Recovery

Pulse sequence used is 180_o 90_o - tR -180° as shown in

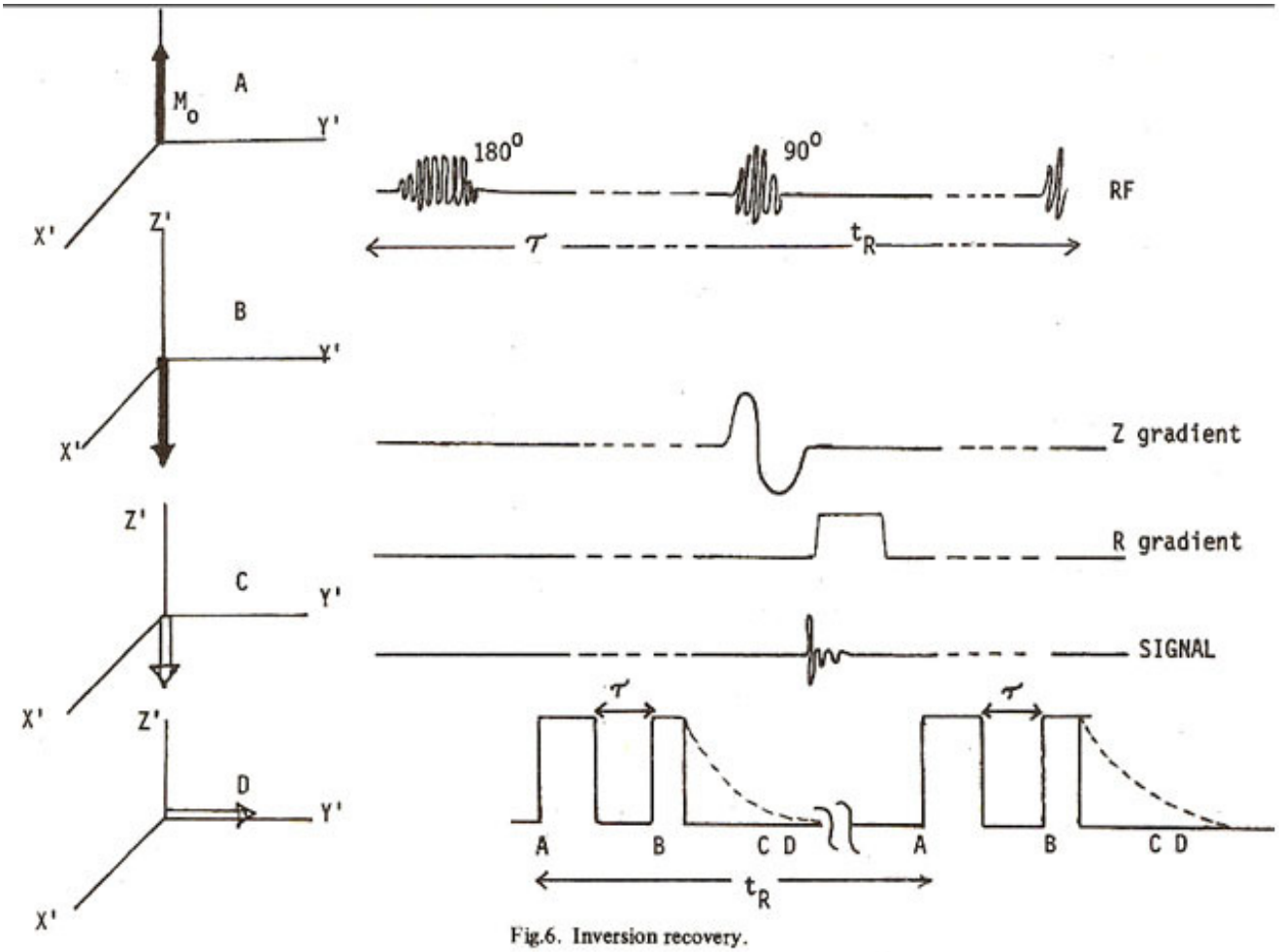

Fig.6. Inversion recovery.

Fig. 6. A 180_o RF pulse is applied to rotate the magnetization M_o in the Opposite direction. Magnetization relaxes following 180_o pulse, back towards its equilibrium position by an exponential process with a rate constant T_1 . r is the time taken by the magnetiza. tion to relax. After a certain time r, the sample is subjected to a 900 pulse which rotates the remaining longitudinal magnetization on to the Y' axis; a FID results which is sampled to generate NMR T_1 signal. When 'r- ln $2T_1$ no signal will appear as at this time Mz is zero. For longer values of r larger and larger signals are observed. After a time tR provided $^tR >> T_1$, the whole sequence is repeated. The height of the F1D signal is measured by sampling the FD at a certain delay after the 90_o pulse in the sequence.

SPIN ECHO. Spin echo sequence (Fig.7)

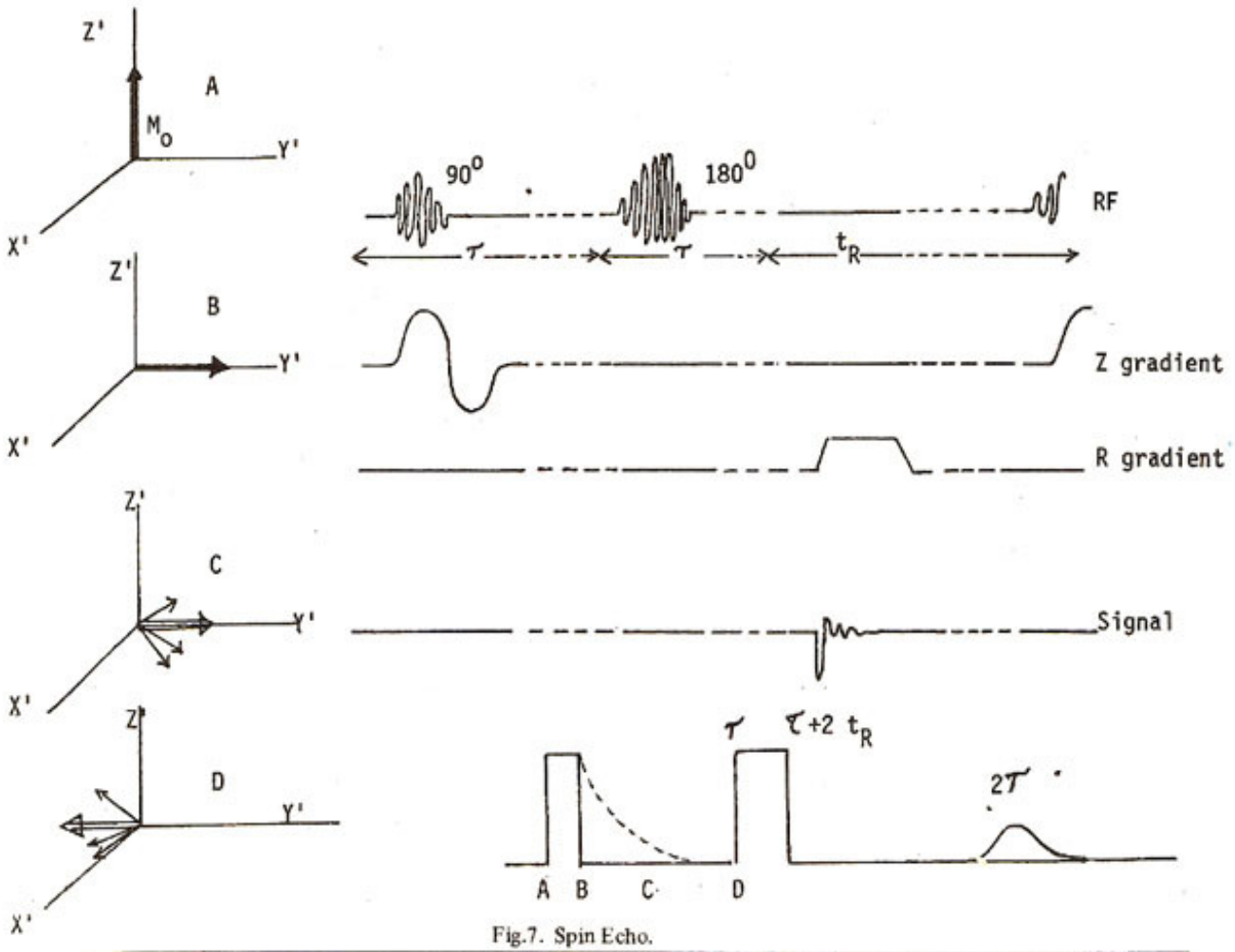

Fig.7. Spin Echo.

produce a signal strongly dependent on T2. RF pulses applied in this sequence is 90_ox' r -180_oy' . 2r - 180_oy' . 2r . 180_oy'.The phase coherence is lost after the first echo, however, it refocussed by the application of 1800 pulse r s later. Thus if a sequence of pulses at times (2n + I) r are subjected, a train of echoes commonly known as Carr-Purcell-Meiboon-Gill chain at time (2n+2) r are formed (Fig. 8).

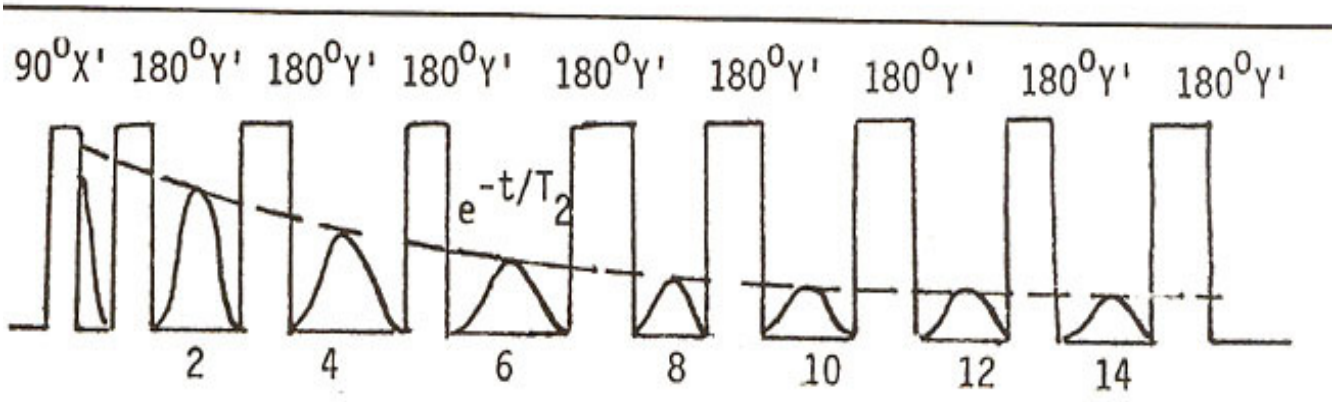

Fig.8. Carr-Purcell-Meiboo n-Gill(CPMG) sequence.

T_2 is obtained by determining the height of echo train decays provided that r should be kept as short as possible so that significant diffusion does not occur between pulses as this would prevent echo formation. may be excited by use of two circular coils of wire with reversing current through each coil.

The magnetic field produced is equal to the math field value H0 only at the midline and nowhere else. In selective irradiation, specific nuclei of the sample become excited by the modified frequencies of the specially formed RF pulse which consist of only a very narrow band of frequencies, and only those nuclei within a plane perpendicular to the direction of the gradient will be excited, since nuclei in this region alone are situated within a magnetic field corresponding exactly to the pulse resonance frequency. The thickness or position of the plane can be altered by changing the width or the frequency offset of the irradiated spectrum electronically.

In an oscillating field, the current of the gradient coil changes periodically. Magnetic field time invariance occurs in but one site between the coils of reversing current. Outside this plane, signals from the nuclei contribute nothing to the image formation.

NMR Imaging

NMR imaging data can be abstracted from NMR signals by single point data acquisition, by line, by plane, or by three dimensional volume. Point or line scanning is done by moving point or line electrically through a plane during data collection and second dimension reconstruction. Volume data acquisition is the most efficient scanning procedure. These data can be reconstructed three dimensionally. Most NMR scanners employ the plane selection method discussed below.

The two commonest methods of plane selection techniques are; 1)oscillating magnetic field gradient and 2) selective irradiation. In either method, the required plane is subjected to a 90_o magnetic field gradient. The field gradient

Mapping the distribution of' nuclei of a selected plane by the reconstruction process takes place through subjecting of the plane to a magnetic field gradient; one side of the plane, more than the other, will respond markedly to the magnetic field. Thereupon an excited RF pulse is applied to the whole plane, and the NMR signal is recorded. The nuclei within the plane then emit NMR signal-nuclei in the low strength portion of the field will emit low frequencies, and high strength portion nuclei will emit high frequencies. The emitted signals yields a frequency spectrum resolving the signal strengths of each frequency. The location of the nuclei emitting signals of a given frequency has not been determined yet. Nuclei may dwell permanently somewhere along a line of identical field strength within the plane; the spectrum give rise to one dimensional projection of the plane under imaging. An NMR scanner now rotates the direction of the gradient, producing a signal that constitutes a new projection, in the same way a CT scanner rotates the direction of the translating beam. A computer at this stage rearranges these projections and reconstructs graphically an image of the plane.

Plannar or whole volume techniques are generally used in NMR imaging due to its enhance sensitivity. Whole volume procedures have some technical constraints e.g. the turn over of large data required large computer memory capacity. In whole volume three dimensional imaging, 256 data points in each dimension with 256 levels of signal intensity are generated for each data point. This requires a computer memory of 134 million bits (256^3 x 8). The data collection time particularly of T_1 maps are increased due to many sequential values of field gradient magnitude or direction to be needed to encode all data the data points in a three dimension matrix. In view of this difficulty, it is preferable to generate a small number of selected two dimensional images. This takes less imaging time than three dimension image. The three dimension image can be dissected at lessiure into great many slices, so that the imaging per plane is reduced.

Resolution, Imaging Time and Signal to Noise Ratio

At present NMR system can achieve a resolution similar to that of CT x-rays. The resolution and contrast sensitivity of NMR depends on the object size, because noise and the potential for distortions increases with the size of the object placed in the RF coils. The resolution is limited due to the detection of voltages induced in the receiver coil by the nuclei. The limit is reached when the signals from the decreasing volume under study are observed by the noise generated within the object and the apparatus.

Technically speaking, resolution in three dimension will be improved by a factor of 2 provided each voxel is reduced in volume by a factor of 8. If all other parameters are constant, the time required to obtain an NMR signal of the same S/N ratio is increased by a factor of 64. On other hand, the imaging time is increased by n^6 If resolution alone improves while the thickness remains constant, the imaging time increases only by n^4. In either case, imaging time for improved resolution is affected. However, for long observation periods the NMR S/N ratio improves to the square root of the imaging time. In short, resolution, S/N ratio, and imaging time, are interrelated in the same way as that of CT x-rays. However, NMR imaging time should be kept as short as possible without sacrificing S/N ratio or anatomic resolution.

At present, the best proton whole body NMR tomographic images of 10 mm slice thickness use sequential plane methods in the frequency range of 2 - 6.5 MHz, giving a resolution of about 2 mm with imaging time of 2 minutes.

Clinical Applications

Physicians generally agree that chemical and physiological changes should follow alterations in histology or anatomy. NMR responds positively to this need by monitoring the biochemical differences in normal structures of a human body. It is interesting to note that the relaxation times of malignant and normal tissues are different; this pecularity may provide a clue for screening malignant disease.

X-ray computed tomography (CT) provide images of normal and abnormal intracranial tissues. Transmission CT image contrast depends on the specific gravity between normal and abnormal tissues. With NMR, brain tissue divides into two categories, gray matter and white matter. Gray matter appears as a collection of nerve cell bodies, and white matter consists of nerve fibers. X.rays CT does not differentiate between gray and white matter. Gray and white matter can be distinguished by NMR, because the hydrogen in gray matter is nearly all water, whereas in white matter a significant amount is contained in lipids.

A vivid contrast can be seen between gray and white matter in NMR images, and thus make it plausible to define internal structures with in the brain with a clarity not seen with CT. This principle is used to visualize tumors within an organ. Tumor cells contain 20% more water than normal cells. Due to higher percentage of water in tumor cells, contrast between normal and abnormal cellular regions appear moredramatically and pronounced than any other imaging modality todate.

The difficulty of bones imaging stem from beam hardening of x-rays in CT; because of this difficulty certain regions in the posterior fossa and other regions of the nervous systems surrounded by thick bone layers are not well highlighted in contrast and resolution. NMR on the other hand attemçted and succeeded in imaging such as pontine bone nuclei, the thickest and densest bones in the head.

The çrognosis and therapeutic handling of patients with known malignant diseases is usually dependent on the speed of the growth of a tumor. CT imaging has been used successfully in the prognosis of disease. NMR is believed to function like CT without ionizing radiation. NMR imaging may be able to make specific diagnosis of tissue cancer, based on the chemical composition of the malignant tumor in its early stages without known evidence of neoplastic disease. This fact is substantiated clearly by the location of a sarcoma implanted in the abdomen of a laboratory animal and consequent depiction by NMR imaging. Further work in defining NIMR parameters for normal and abnormal tissue and in the definition of image contrast between them needs to be further evaluated before these images can become clinically useful.

The most promising aspect of NMR imaging other than proton in the human body is phosphorous-3 1. Phosphorous compounds occur naturally in the form of phospholipids in membrane structure and other tissues like nervous tissues, muscles, and liver. Concentration of P.31 in different brain regions would be of great value in understanding brain function.

P-31 as we all know, is the essential ingradient of the high energy molecules adenonsine triphosphate (ATP) and phosphocreatine which involves in the process of energy transfer in the living cells.

Turnover of P-3 1 in the human body reveals several NMR spectra shifted resonance peaks whose heights corresponds to the concentration of the individual phosphorous compound. Turnover of specific phosphorous compound in a certain organ may give an indication on the metabolic status. The tumors metabolic characteristics, both in the initial stage and later stages of growth as it becomes more hypoxic, may provide a clue to differentiate between neoplastic tissues from surrounding aerobic tissues. It is however, not clear if the spectral changes in this case, can distinguish between benign and malignant tumors or from those of nonmalignant abscesses? It is therefore, essential to study each tumor type and its counterparts as well, before these spectral changes can clinically feasible in the process of monitoring metabolic rate.

Preliminary results in murine tumor models from a number of studies shows that P-31 NMR spectroscopy can monitor the metabolism rates in tumors during untreated growth and can detect tumors response to different therapeutic -modalities (Evanochko et al., 1983).

In the presence of active hepatitis and early cirrhosis of the liver, the relaxation time (T_1) increases. Depending on the severity of the disease, the areas of increased T_1 appears to spread from around portal tracts. Comparison of NMR (T_1) images with Tc-99m sulfur colloid scan of liver, it has been found that NMR is more sensitive than the nuclear medicine procedure.

With NMR visualization of kidney, there is good differentiation between the cortex and medulla, and the columns of Bertin can be seen clearly although diaphragmatic movement may blur their contrast. The impact of NMR imaging and clinical efficacy will determine the advantages and limitations in comparison to other imaging modalities in years to come.

NMR Contrast substances

Contrasting substances alter relaxation times (Dikson and Ekstrand, 1982). By virtue of this change, uptake, turnover, and clearance (washout) of the tracer can be observed in the same way as used in nuclear medicine radiopharmaceutical studies. In conventional x-rays imaging barium and iodine are used to enhance contrast of an organ. In order to use the same concept with NMR, paramagnetic substances are used o differentiate morphology and to measure physiologic function. The prevailing contrast agents are Mn++ and Fe+++. They have however, been determined to be toxic in the human body. The search for suitable non toxic paramagnetic substance is going on to enhance future NMR imaging.

The presence of hydrogen proton in the human body are enormous and this abundance makes the present NMR imaging technique to evaluate individual details impossible. In order to extract more elobrate diagnostic informtion, fluorine in the organic molecules @erfluorocarbon compounds) which is used in making synthetic blood and can be used as a contrast agent (Peter sons, 1983). The advantage of using fluorine is that human body contain very small amount of this compound, and when this artificial blood is subjected to strong magnetic field. it emits a strong NMR signal. It has been proposed that this synthetic blood in small doses, would enable to visualize the blood vessels and blood pool flow through out the body. However, it remains to be seen whether it is possible to do pulmonary and brain perfusion scans. Again the question that has to be answered are: how safe is this compound and what is the biological half of this compound in the body. Once these questions have been resolved, then the NMR imaging will open a new vista of prognosis.

Hazards in NMR Imaging. There are three potential sources of hazards in NMR imaging:

1. Exposure to the static magnetic field.
2. Exposure to rapidly changing magnetic field.
3. Heating due to RF power.

No biological effect has been reported. However, apprehension of risk has been shown in conditions such as cardiac arrhythmias, pregnancy, epilepsy. Caicraft (1983) drew attention on a plausible risk and hazard. He stated: “patients should be checked before being imaged for pace makers, metal objects and metal surgical implants. The chief hazard to a patient seems to be a pair of scissors escaping from a

pocket as the nurse leans over to position them and heading past at a into the machinel" LIMITATION- Due to large imaging time, the reproduction of NMR signals in the case of a restless patient is a major problem. Due to this longer exposure, several kinds of artefacts will appear on the image and make the prognosis a difficult proposition. Another limitation, analogous to beam hardening in x-rays CT, is the attenuation of the RF and non-uniformities of the magnetic fields.

Conclusion

NMR medical imaging technique in a short period of time has made a spectacular impact on the prognosis of benign and malignant diseases. With the first generation NMR system currently in use, it is obvious that the improvements in NMR imaging in future are likely to be made in increasing the spatial resolution reducing magnetic field strength and radiofrequency power. Sequences have to be improved, new contrast substances to be developed and scanning time to be shortened etc. The advantage of NMR imaging that it allows variety of sequences to be used for pictorial representation of anatomy and physiology of the human body, and this is its unique capability. At present only small domain of NMR's application in medicine have been exploited, though in the next few years, with all the limitations of first generation NMR system resolved and improved, will explode on medical scene as an indespensible imaging modality.

References

1. Apparatus and method for detecting cancer in tissue. US patent No. 3789832 filed March 17, 1972.
2. Bloch, F., Hansen, W.W. and Packard, M. (1946) The nuclear induction experiment. Phys. Rev., 69 : 127.
3. Caicraft, M.E. (1983) New. view into the body. Phys. Bull., 34 : 313.
4. Damadian, R. (1971) Tumor detection by nuclear magnetic resonance. Science, 171 : 1151.
5. Dixon, R.L. and Ekstrand, K.E. (1982) The physics of proton'NMR. Med. Phys., 9: 807.
6. Evanochko, W.T., Ng, T.C., Lilly, M.B., Lawson, A.J., Corbett, 1.11., Durant, i.R. and Glickson, J.D. (1983) In vivo 3lp NMR study of the metabolism of murine mammary 16/C adenocarcinoma and its response to chemotherapy, x-radiation, and hyperthermia. Proc. Nati. Acad. Sci. USA, 80: 334.
7. Gore, IC., Emery, E.W., Orr, J.S. and Doyle, F.H. (1981) Medical nuclear magnetic resonance imaging. I. Physical principles. Invest. Radiol., 16 : 269,
8. Hinshaw, W.S. and Lent, A.H. (1983) An introduction to NMR Imaging; from the Bloch equation to the imaging equation. Proc. IEEE, 71: 338 - 350.
9. Holland, G.N. System engineering and instrumentation of a whole body proton NMR imaging system in nuclear magnetic resonance (NMR) imaging. Edited by Partain, C.L. et al. Philadelphia, Saunders, 1983, p.130.
10. Lauterbur, P.c. (1973) Image formation by induced local interactions; examples employing nuclear magnetic resonance. Nature, 242: 190.
11. Lerski, R.A. (1983) Physical principles of nuclear magnetic resonance imaging. Radiography, 49: 85-90.
12. Partain, c.L, Price, R.R., Patton, J.A., Stephen, W.H., Stewart, R.G. and James, A.E. The physical basis of NMR imaging, in nuclear magnetic resonance (NMR)imaging. Edited by Partain, ci. et al.
13. Philadelphia, Saunders, 1983, p. 91.
14. Pennock, J.M. (1982) Nuclear magnetic resonance imaging at Hammersmith hospital. Radiography, 48 :221.
15. Petersons, 1. (1983) Artificial blood and NMR imaging. Sd. News ,123 : 381.
16. Purcell, E.M., Torrey, H.C. and Pound, R.V. (1946) Resonance absorption by nuclear magnetic

moments in a solid. Phys. Rev., 63:37.
17. Pykett, I.L. (1982) NMR imaging in medicine. Sd. Am., 246: 78.
18. Pykett, I.L., Newhouse, J.H., Buonanno, F.S., Brady, T.J., Goldman, MR., Kistler, J.P. and Pohost, G.M.(1982) Principles of nuclear magnetic resonance imaging. Radiology, 143:157.