MAGNETIC RESONANCE IN MEDICINE AND POSSIBILITIES IN TURKEY

by

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- To My Parents -

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ABSTRACT

Magnetic Resonance is a new diagnostic imaging method in medicine, although it has been used for a long time in other fields, including biology, chemistry and biochemistry.

Magnetic Resonance Imaging or shortly MRI is based on the different behaviours of various atomic nuclei in the human body. A static magnetic field and a changing radio-frequency field are applied to the body of the patient. By using the echo signal from the tissues and processing it properly one can obtain the image of the body on agiven plane.

Due to the metabolic structure of the tissues, different signals are obtained which are dependent on certain parameters like relaxation times T1 and T2. These acquired signals are then processed and developed to MR images by using different imaging methods.

MR offers very good images with a very high resolution and the possibility of direct imaging from transaxial, coronal and sagittal planes which are not easily achieved in other imaging techniques.

MR replaces Computed Tomography, conventional X-ray, nuclear imaging methods and others in many cases. Furthermore,the patient is fortunately protected from hazardous effects of those examination techniques.

The installation of an MR system brings some difficulties that don't appear for other medical diagnosis systems. The solutions to these problems, however, are available to a great extent.

MR can be made economically feasible although it has a fairly great over-all cost including the capital and operational expenses.

In this thesis MR imaging is studied from different points of view. Its physical principles are given. Different measurement and image reconstruction techniques are discussed. The known medical applications of MR are also listed. A comparison between MR and other diagnostic imaging modalities is done.

Finally a case study involving an eventual installation of an MR system in Turkey is presented. The feasibility of such aproject is discussed.

ÖZET

Magnetıc Resonance, bioloji, kimya, biokimya gibi alanlarda uzun süredir kullanılmasına karşın, tıp için oldukça yeni bir görüntülü teşhis metodu.

Magnetıc Resonance sisteminde görüntü elde edilmesi, insan vücudundaki atom çekirdeklerinin değişik davranış şekilleri göstermeleri esasına dayanıyor. Sabit bir manyetik alan ve değişken bir radyo-frekans(RF) alanı hastanın vücuduna uygulanıyor. Vücuttaki çeşitli hücrelerden yankılanan sinyallerin işlenmesi ile de arzulanan bir düzlemde görüntü elde ediliyor.

Ölçme sonucunda insan hücrelerinin metabolik yapılarına bağlı olarak değişik sinyaller elde ediliyor. Normale dönme zamanı veya relaksasyon zamanı olarak isimlendirebileceğimiz T1 ve T2 gibi parametreler, sinyallerin karakterleri üzerinde etken oluyor. Bu sinyaller işlenerek MR görüntülerine dönüştürülüyor.

MR yüksek rezolüsyonlu çok net görüntüler elde edilmesini sağlıyor. Ayrica diğer metodlarla direkt olarak elde edilmesi zor olan koronal, sagıttal ve transaksiyal düzlemlerde görüntüleme de MR'ın getirdiği imkanlar arasında.

MR pek çok vakada bilgisayarlı tomografi,konvensiyonel röntgen, nuklear teşhis cihazları ve diğer sistemlerin yerini dolduruyor.

Bir MR sisteminin montajı ve yerleştirilmesi diğer cihazlarda rastlanmayan bazı zorluklar getirebiliyor. Ancak bu konuda pek çok çalışmalar yapılıp, zorluklar büyük ölçüde giderilmiş durumda.

MR'ın satış fiatı ve kullanım giderleri çok yüksek görünmesine rağmen ekonomik açıdan uygulanabilir bir sistem.

Bu tezde, MR çeşitli yönlerden inceleniyor. Fiziki prensipleri tanıtılıyor, ve değişik ölçme ve görüntüleme teknikleri açıklanıyor. MR'ın tıptaki kullanım alanlarından söz ediliyor. MR ile bilgisayarlı tomografi, konvensiyonel röntgen, nuklear tanı yöntemleri ve ultrasonografi karşılaştırılıyor.

Son olarak da Türkiye'de bir MR sisteminin kurulabilirliği çeşitli yönlerden tartışılıyor.

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1. INTRODUCTION

To acquire images from inside the human body has been a dream for years until the discovery of X-ray in 1895 by Conrad Rontgen. Although other diagnostic imaging techniques, including ultrasound devices and nuclear cameras have been developed in the mean time, radiographic methods using X-rays have been accepted as one of the best techniques in imaging. Especially with the development of Computed Tomography (CT) scanning, a milestone in diagnostic X-ray history, extremely good images can be obtained.

However, X-rays have one significant disadvantage- they are an ionizing radiation and so can do harm in the courses of doing good. A similar kind of disadvantage also applies to interior imaging using gamma-rays. There are other imaging methods like ultasound, but unfortunately they do not always lead to a diagnosis.

A new diagnostic imaging method, called "Magnetic Resonance" or shortly "MR" has overcome these disadvantages by using high magnetic fields to obtain high contrast cross-sectional images of any desired plane of the human body. Great advances have been made in recent years in the development of MR imaging. A great number of Whole Body MR instruments have been installed and are currently undergoing clinical evaluation.

In this thesis, an overwiew of Magnetic Resonance Systems including physical basis, imaging characteristics, medical applications, installation problems, advantages and disadvantages with respect to other imaging systems will be provided. Considering the conditions for Turkey, the possibility of an MR installation in a turkish medical institution will be discussed.

After this introductory presentation about MR, in Chapter 2 some comments will be made on the history and development of Magnetic Resonance in medicine and in other fields of science, including chemistry, biology and biochemistry.

In Chapter 3, the physical principles on which the magnetic resonance is based will be discussed. The characteristics of atomic nuclei which play an important role in the whole phenomenon and also the parameters of MR will be further discussed.

Chapter 4 is about the MR imaging where imaging methods and measurement techniques of different kinds are explained in detail. Also some techniques, like sodium or phosphor imaging which cannot be applied yet in routine diagnostic will be important in a near future are mentioned in this chapter.

'Construction of an MR system' is the title of the 5th Chapter. All the necessary equipment and instrumentation that built up a magnetic resonance device will be mentioned and explained comprehensively. The magnet, being the heart of the whole system has the

largest place in this chapter.

Chapter 6 is about the application possibilities of MR in medical diagnosis. Comparison of MR with other diagnostic methods, advantages and disadvantages, possible side effects are some of the topics discussed in this chapter. Also pictures of acquired images are presented here.

Installation diffuculties, reasons of their apperances, possible solutions, and choices that can be met for installation are given in Chapter 7. Installation of an MR system is in reality more difficult than other imaging equipment, but nearly ideal solutions have been almost developed and they will be provided in this chapter.

Chapter 8 has the topic 'financial aspects of an MR system' and the costs of such a system are discussed here. The first prerequisite of the feasibility of any investment is a realizable position of its break-even point. Such an analysis is done for an MR device in Chapter 8.

Chapter 9 discusses the possibility of MR in Turkey. There is not yet any MR system in Turkey. It is very often discussed whether it is the right time now to purchase Magnetic Resonance for a medical institution in Turkey. People are afraid of getting

many technical and economical problems with an MR because it is very new and not well known. The aim of this chapter is to clear some confusions in minds and give an example to the financial calculations of an MR installation.

Chapter 10 is the conclusion of this thesis.

2.HISTORICAL PERSPECTIVE OF MR

The use of Magnetic Resonance in medical applications did not start until recently. But in reality, MR spectroscopy has existed since 1945. It has been used almost entirely by physicists, chemists, biologists, and biochemists to study atomic and molecular structure. MR has played an important role in the development of the theories of solids and liquids and has provided a fertile test bed for spectroscopic and quantum mechanical theories.

MR scanning for medical diagnosis was first reported by Damadian in 1971 followed by others. In 1973 Lauterbur proposed the use of MR as a method of obtaining images of hydrogen density in the inside of objects. His proposed technique used the same type of reconstructions as computerized tomography and thus was subject to similar artifact problems.

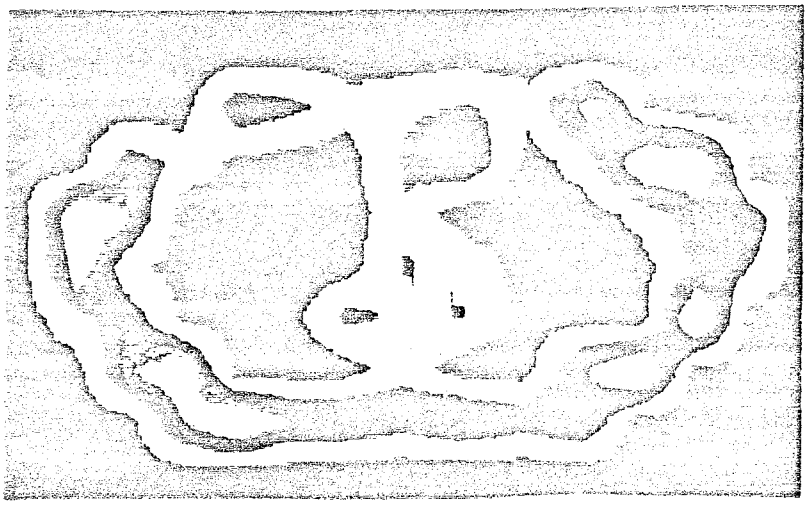
The use of electronic scanning did eliminate moving parts and, as with all MR techniques, there was no exposure to ionizing radiation. Other imaging techniques, including selective point,sequential line, selective irradiation, multidimensional Fourier transforms have been developed by Hinshaw, Kumar and others.

The practical realization of the first MR body scanning machine was made possible in 1977. It was constructed at the Downstate Medical Center, Brooklyn,

New York. This MR device, called 'Indomitable', was a 0.5 T liquid helium cooled superconductive magnet and a computer driven scanning bench, developed by Dr.Minkoff, Dr.Goldsmith, and Dr.Damadian.

The first attempt to achieve a live human scan with this machine was made on May,1977. Baseline blood pressure, respiratory rate, pulse rate, and electrocardiographic determinations exhibited no significant changes. The scan, however, failed due to excessive loading of the antenna by a sample too large for the dimensions of the RF coil.A second attempt on July 3,1977 gave better results. The scan continued approximately 5 hours and resulted in a respectable image of the full-sized human torso(see Fig.2.1.). For that scanning the operation frequency was 2.18 MHz, the magnetic field was 508 Gauss.

Figure 2.1. The first human scan obtained July 3,1977. The image is a cross-section through the chest at the level of the eighth thoracic vertebra. The image shows the body wall, the right and left lung field, the cardiac chambers, right attrium and a ventricle.



The successful experiment with the first MR device led other groups to carry out researches into MR technique. There were technical problems in the realization of advanced MR devices, including the limitations in magnet technology and RF coil construction. It took some years to bring solutions to these problems.

Today, the development of MR systems have reached such a stage, that they can be set into routine diagnostic medicine. The research works are continued by universities, medical institutions and manufacturers in order to develop advanced techniques. More than 15 companies produce MR systems. General Electric, Technicare, Siemens, Philips, Picker, Fonar,and Elscint are some of these.

The number of MR imaging systems operating worldwide is more than 200. It has been reported that 49 systems have been installed only in the last three months of 1984, an indicator of a considerable increase in the numter of MR systems.

Meanwhile, the name NMR which stands for ' Nuclear Magnetic Resonance (NMR) ' started loosing its wide use and MRI (Magnetic Resonance Imaging) or simply MR became a more popular name. Two reasons helped the word "nuclear" disappear from NMR:

a-) Calling MR 'nuclear' is not completely correct

since chemical shift analysis(explained in section 4.3.5.) involves electrons, and not protons in the nucleus of the atom.

b-) The word 'nuclear' has a bed effect on the public. It conjures up visions of gamma radiation.

The name "MR" started to be accepted by many universities, research institutes, and manufacturers, yet. Therefore, in this thesis the term "Magnetic Resonance(MR)" will be used instead of NMR.

3. PHYSICAL PRINCIPLES OF MAGNETIC RESONANCE

The physical principles of Magnetic Resonance was recognized and described in 1946 by the physicist Bloch and Purcell approximately at the same time.For a better understanding of the principle of an MR system, the behaviour of nuclei in a strong magnetic field must be studied.

3.1. Characteristics of the Atomic Nucleus

Atomic nuclei consist of subatomic particles, called neutrons and protons. Beside the characteristic of being electrically charged, these nuclei also have a property, called " spin " . The single nucleus may or may not have a spin depending upon the total number of particles, or more precisely whether this number is odd or even.

Since nuclei are charged particles, their spin is always associated with a magnetic moment. The most frequent nucleus in nature is the proton itself. For instance 1 mm^3 water contains 6.7×10^{19} protons.

Without a magnetic field, the protons are randomly orientated as shown in Fig.3.1. In a magnetic field, however, the protons align with or against this field (Fig.3.2). There is no intermediate state. More protons are aligned with the field than against the field because their energy state is more favorable.

Fig.3.1. Random orientation of the protons in the absence of an external magnetic field

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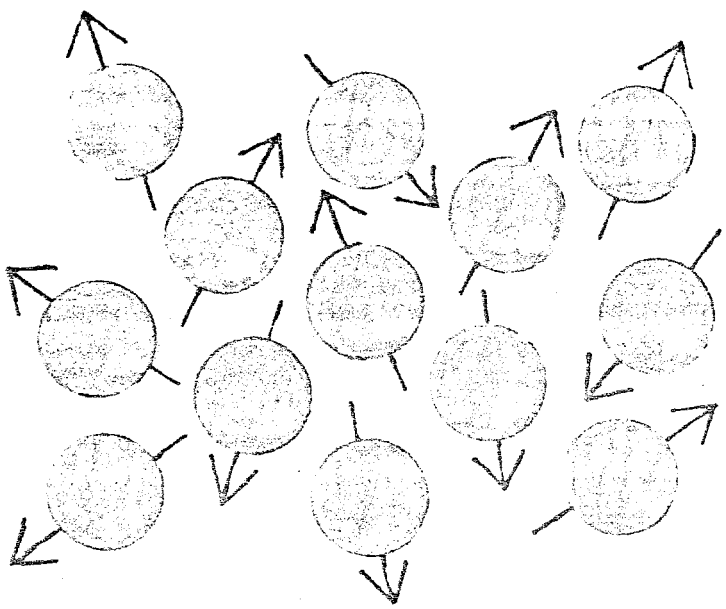
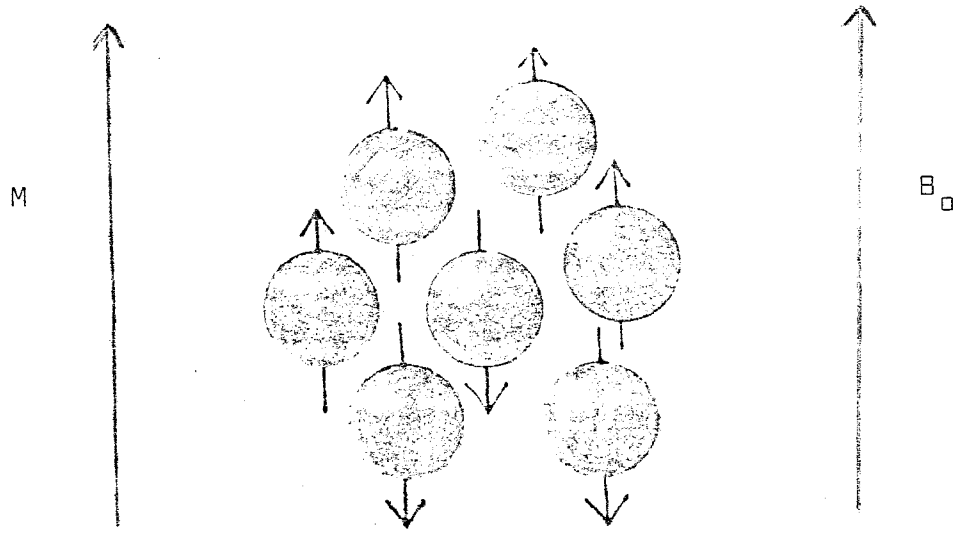


Fig.3.2. Net magnetic moment M aligned parallel to the external field B_0 due to the slight excess of parallel to antiparallel nuclei



The excess of parallel aligned spins yields a nuclear magnetization M of the sample, which is proportional to the magnetic field and inversely proportional to the temperature.

3.1.1. Larmor Frequency

The laws of motion of this magnetization M are symbolically described in Fig.3.3. In the energetically most favorable equilibrium position the magnetization is lined up like a compass needle parallel to the magnetic field. If, however, the magnetization is tilted up by an external perturbation, the angular momentum of all spins prevents the magnetization from turning back to its equilibrium position. Instead it will precess around an axis parallel to the magnetic field like a spinning top in a gravitational field. The angular frequency of this precession, which means the frequency at which the spinning nuclei rotate about their axis, is given as follows:

$$\omega_0 = \gamma B_0 \qquad (1)$$

where γ : gyromagnetic ratio, a constant which is characteristic for the type of nuclei involved

B_0 : Magnetic Field

Fig.3.3 Motion of the nuclear magnetization within B_0

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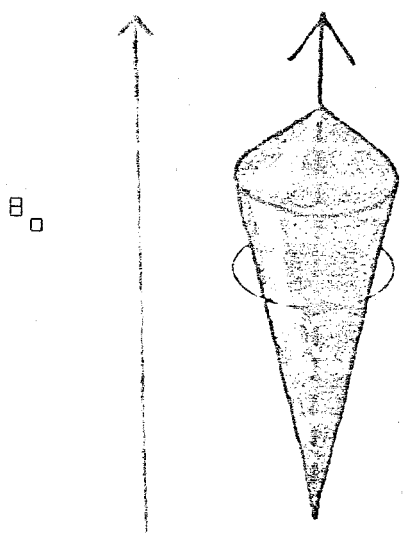
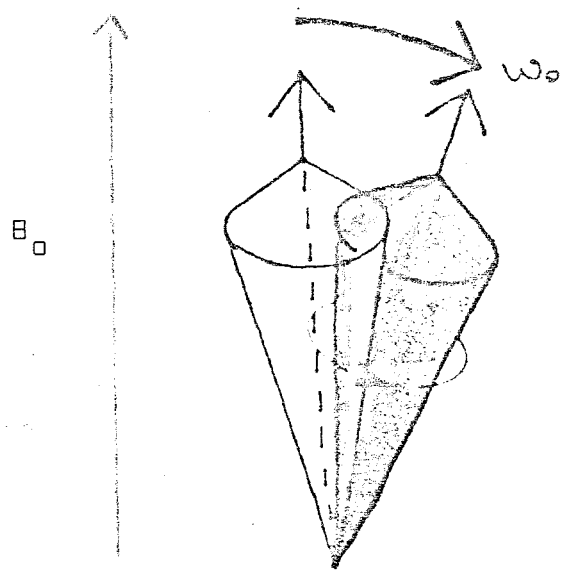
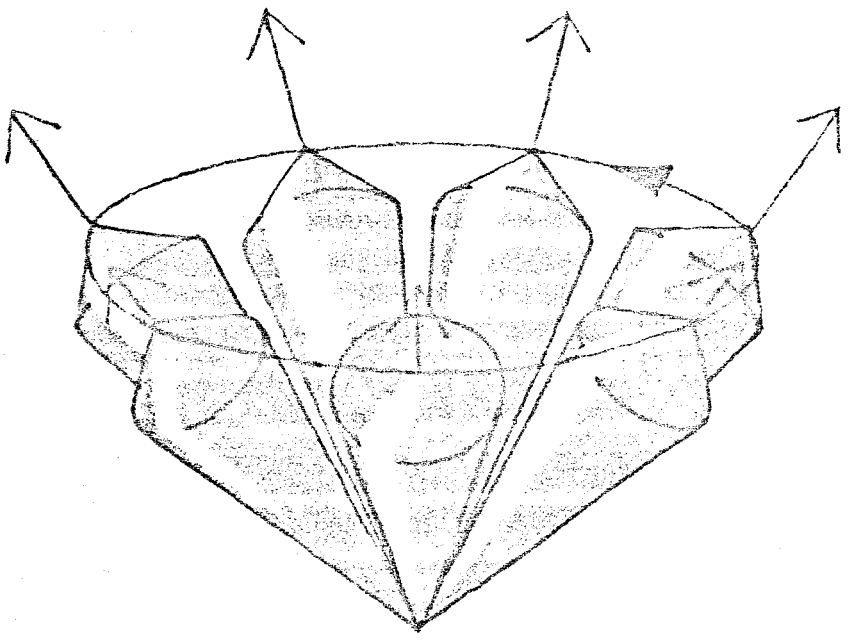
a) In the equlibrium position the magnetization is aligned with the magnetic field



b) After excitation, the magnetization precesses around the field direction



c) Precession of the magnetization in a gravitational field



The quantum description of this same phenomena can also be done. When a nucleus with a magnetic moment is placed in a magnetic field, the energy of the nucleus is split into lower(moment parallel with field) and higher (antiparallel) energy levels. The energy difference between these energy states is such that a proton of a very specific frequency is necessary to excite a nucleus from the lower to the higher state. The excitation energy is dependent on the strength of magnetic field and is determined by the Larmor Frequency,

$$E\ (eV) = \hbar\omega_0 \qquad (2)$$

where $\hbar$ is Planck's constant h, divided by 2π.

Actually, the spins do not just absorb this exact amount of energy. Furthermore, not all the protons have exactly the same resonance conditions. From Equation (1) we can easily compute the frequency needed to observe a resonance in a certain field. For protons in a magnetic field of 1 T, the appropriate frequency is 42.574 MHz, which is conveniently produced by RF techniques. Larmor frequency for other types of nuclei is given in Table 3.1

Table 3.1 Larmor frequency for various nuclei

Nuclei	$f=\omega_0/2\pi$ (MHz/T)
^{1}H	42.57
^{13}C	10.70
^{31}P	17.20
^{15}N	3.10
^{19}F	40.1
^{2}H	6.55

The RF radiation applied to induce the magnetic resonance phenomenon is equivalent to the application of a second magnetic field which is much smaller than the first one and rotates about the net magnetization

vector in the x,y plane. If the frequency of rotation is appropriately chosen, some of the aligned nuclei will experience a torque which will displace the axis of the net nuclear magnetization vector from its position parallel to the static magnetic field (see Fig.3.3b). This displacement causes the magnetization vector to precess like a spinning gyroscope. The angle between the direction of the static magnetic field and the magnetization vector is dependent on the amount of energy delivered by the RF radiation and increases as long as the pulse remains on.

The angle θ can be increased by augmenting either the strength or the duration of the pulse or both. A pulse just long or strong enough to turn the vector from its initial position until it rotates exactly in the x,y plane is called a 90° pulse. If the pulse is turned off, the vector continues to rotate freely inthis plane and generates a signal called the " Free Induction Decay " (FID). The vector then lies wholly in the x,y plane, and the 90° pulse provides a FID of maximum amplitude since it is only the x,y component of the displaced vector that contributes to the emitted signal. A pulse applied for twice as long or with twice the pulse power rotates the vector from its original position by $\theta = 180^{\circ}$, until it points in the diametrically opposite direction; this is a 180° pulse. Since in that case there is no component in the x,y plane, a single, isolated, 180° pulse gives no signal.

When the RF pulse is turned off, the precessional motion of the net magnetization vector decays, the

vector returns to its equilibrium position parallel to the static magnetic field, and the emitted signal diminishes. This phenomenen is called " relaxation ". Relaxation has two variables, namely, relaxation times T1 and T2.

3.2. Relaxation Times

Relaxation times are characterized by two sample related time constants:

a) The longitudinal (spin-lattice) relaxation time T1

b) The transverse (spin-spin) relaxation time T2

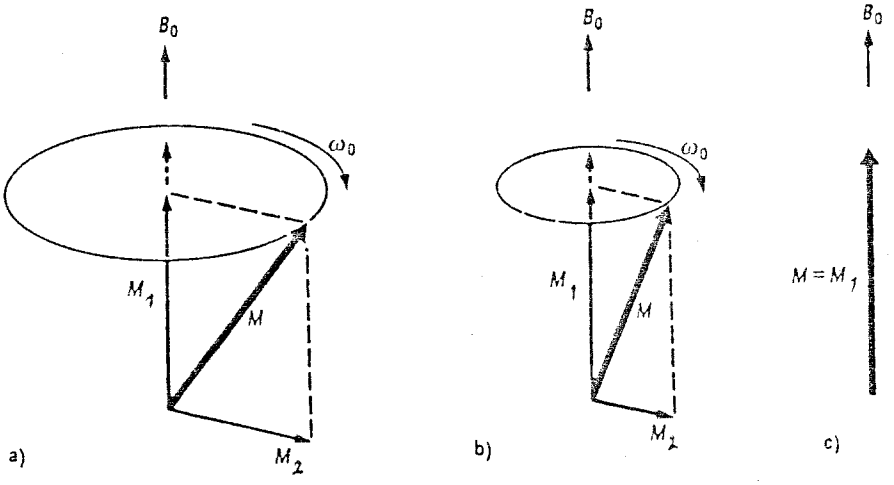
3.2.1. Spin-Lattice Relaxation Time T1

T1 represents the time required for the component of the net magnetization vector in the z-direction to return to its initial value after it has been perturbed by the RF pulse. This reastablishment of bulk magnetization depends on the nuclei losing energy to their environment or ' lattice ', hence the term spin-lattice relaxation. Because the elements of this phenomenon are considered in relation to the original longitudinal orientation (M and B), T1 is also called " The longitudinal relaxation time ".

Spin-lattice relaxation time T1 has a close relation with the spin-spin relaxation time T2. These two competing phenomena can be distinguished as they

can be measured separately. The magnetization vector can be analysed into components M1, parallel to the magnetic field and M2, rotating in a perpendicular plane, as seen in Fig.3.4.

Fig.3.4 Changing process of the precessing magnetization vector from a. via b. to c.



The component of the magnetization parallel to the static magnetic field M1 approaches equilibrium because of the interaction of the nuclear spin system with the surrounding lattice. This can be described mathemati cally by an exponential function

$$M1(t) = M(1 - e^{-t/T1}) \qquad (3)$$

The T1 relaxation process is always an emission of energy, which is equal to the energy difference between the initial and the final energy levels. T1 depends on many parameters, the important ones are as follows:

- Type of the nucleus,
- The resonance frequency and the magnetic field,
- The temperature,
- The presence of macromolecules,
- The presence of paramagnetic ions.

Direct quantitative comparisons between T1 values at different frequencies and so at different magnetic fields is not possible.

3.2.2. Spin – Spin Relaxation Time T2

T2 is the spin-spin or transverse relaxation time. It is a time constant characterizing the return to equilibrium of the component of the magnetization that is perpendicular to the external or static field. It results from the interaction of a spinning nucleus with the spin of an identical nucleus pointing in the opposite direction, hence the name.

As seen in Fig.3.4 ,the relaxation process reduces the transversal component M2 of the nuclear magnetization due to interactions between the neighbouring spins

This is often described by

$$M2(t) = Me^{-t/T2} \qquad (4)$$

After an RF pulse has tipped the nuclear magnetization vector towards the transverse plane, the components of this vector all precess in phase- they are coherent. However, each proton is not only influenced by the static magnetic field but also by the protons in the neighbourhood. This fact causes subtle local alterations and so some nuclei precess at different rates from others, the RF waves from individual nuclei dephase and cancel each other out, and the sum of nuclear magnetization vectors in the transverse plane decays to zero. the overall time characterizing this dephasing or decay is called T2'. The observed parameter T2' is in fact a composite relaxation time resulting both from intrinsic interactions between neighbouring nuclei and from heterogeneites in the applied magnetic field. For an exactly homogeneous magnetic field, we measure not T2' but only T2. We call T2 as ideal spin-spin relaxation time and T2' as effective spin-spin relaxation time.

Spin-spin relaxation time T2 depends on a number of parameters:

- Observation frequency,
- The temperature,
- The movability of the observed spins,
- The presence of macromolecules and paramagnetic ions.

In fluids, T2 is nearly equal to T1. In solids both T1 and T2 are most sensitive to the degree of molecular motion. In solids and at low temperature in other stages, there is little motion and T1 may be many seconds while T2 is only microseconds. However, in liquids and at higher temperatures in other stages, T1 and T2 are almost equal, both being about two seconds for pure water. Therefore, if the ratio of T2 to T1 approaches to 1, the sample may be assumed to be relatively "liquid-like",and if the ratio is very small, the sample is "solid-like". That means, the object to be measured is:

solid or soft tissue if $T1 >> T2$, and

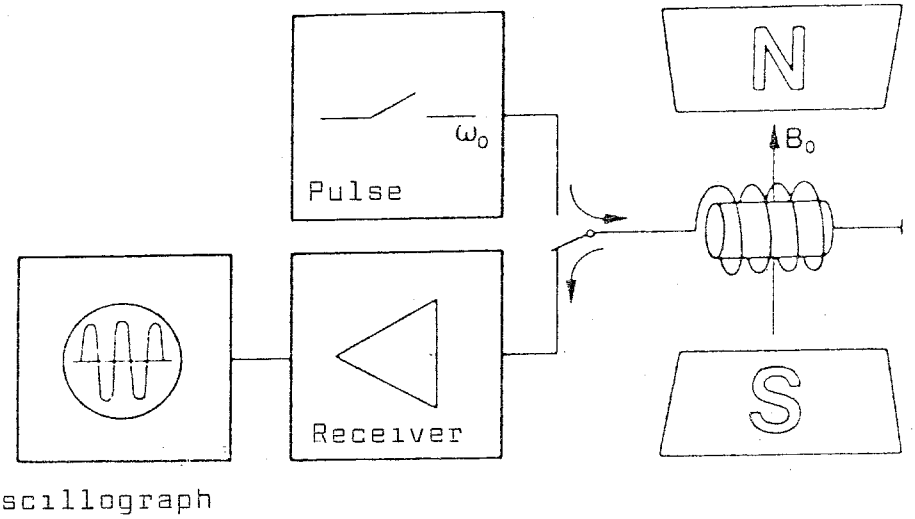
liquid if $T1 \cong T2$

In MR imaging, only the signal from "liquid-like" regions is observed, rigidly bound nuclei give essentially zero signal. Variations in T1 proton relaxation time among different tissues are often related to free water content.

4. MAGNETIC RESONANCE IMAGING (MRI)

For chemical analysis of homogeneous samples, it is sufficient to surround the glass case where the sample stays with a high frequency coil and process with the apparatus seen in Fig. 4.1.(105)

Fig 4.1. Basic MR experiment with high frequency pulse excitation



For in-vivo applications of MR in medicine, however, a spatial relation is necessary because of

complex structure of human organs.

Since we have to think that every picture consists of single picture elements, which are called " pixels " it is necessary to win the MR signal out of various regions of the body gradually or simultaneously.

4.1. Basic Ideas of MRI

Magnetic Resonance Systens that are being used currently for routine medical diagnostic purposes, base on the measurement of the MR signal of the cell water in the human body.The reason is that the acquisition of images from hydrogen atoms is easier than acquiring from other molecules because of the following two reasons :

a-) High MR sensitivity of H-atoms,
b-) High water concentration of human cells, almost 75%

There are also imaging methods that use phosphor, sodium, and fluorine for acquisitior purposes, but they are still used for experimental studies. These methods will be discussed in Section 4.5.

4.2. Imaging Methods

For magnetic resonance imaging it is necessary to differentiate the contributions to the MR signal from the various regions of the body being examined. For

this purpose, a number of procedures have been developed which are based on the same fundamental principle.

The strength of the static magnetic field is made dependent on the location in a specific manner. Because the resonance frequency and the magnetic field strength are proportional to each other, it is possible to allocate the various contributions to the MR signal to the site of their origin, by means of the frequency. The principle of topological MR is a simple procedure to achieve a location-resolving measurement, only the MR signal from a limited volume element (voxel) of the object being examined is detected. By the successive measurement of a series of voxels in the human body, the image of a slice can be built up. However, much more rational methods are those which record the MR signal simultaneously from larger regions of the object being examined or, in certain circumstances, from the whole object and by applying suitable reconstruction methods, to form an image of the object from this data.

These methods are differentiated by the size of the partial volume which the MR signal is simultaneously recorded and evaluated. There are following four categories, which are shown in Fig.4.2.(105)

a-) Sequential point imaging,
Sensitive point method
Focused nuclear resonance

b-) Sequential line imaging,
Selective excitation line scan
Sensitive line or multiple sensitive point

c-) Sequential plane imaging,

2D projection reconstruction

2D Fourier imaging

Planar imaging

2D Echo planar imaging

Rotating frame imaging

d-) Three dimensional imaging(Simultaneous methods)

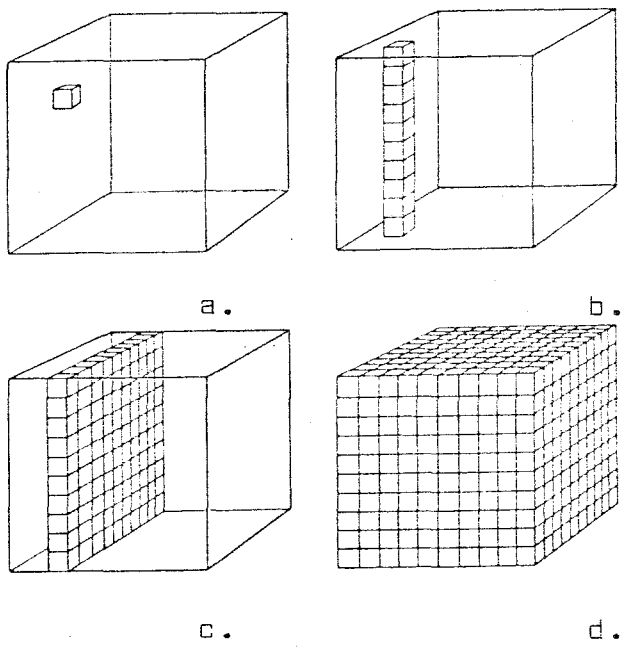
3D Projection reconstruction

3D Fourier imaging

Multiplanar imaging

3D Echo planar imaging

Fig.4.2. Imaging methods that are used in MRI



A successful technique used in all reconstructional imaging methods to obtain spatial information regarding the distribution of magnetic nuclei within an object is superimposing over the static B field another electromagnetic field which varies linearly across the region of interest. In this circumstance, each plane of an object oriented perpendicular to the gradient direction will experience a different B field. Thus each plane will resonate at unlike frequencies depending upon their location. Recalling the fact that the Larmor frequency is not only dependent upon the characteristics of the nucleus but also upon the environment B, known correspondence between the resonance frequency and the spatial location in the direction or plane of interest is established. The MR signal at each frequency thus becomes a one dimensional projection (line, plane, or volume) of the spatial distribution of magnetic nuclei within the object that is studied. By rotating the direction of the gradient relative to the object, additional projections can be obtained. Using techniques of reconstruction from projections similar to those used for CT imaging, two dimensional images are produced.

4.2.1. Sequential point imaging

In sequential point imaging methods the MR signal at each instant comes from one small volume element in the sample. The simplest method of obtaining an MR signal from a localized region is to use a small flat receiver coil positioned over the region of interest.

selectivity of such a surface coil is approximately confined to a volume subtended by the coil circumference and one radius deep from the coil center. Some variations in depth is possible by changing the MR pulse lengths, but scanning the localized volume typically requires manually moving the coil and retuning it. For these reasons the method is not suitable for generating an image of the entire sample in three dimensions.

In Fonar method, signal from the sample is achieved by suitable shaping of the static magnetic field after recording the signal from one voxel. The experiment is repeated traversing the volume element sequentially through a defined plane in the object, mapping the MR signal throughout the plane.(9)

"Sensitive Point Method" of Hinshaw is the most sophisticated sequential point imaging methods, requiring no moving coils or gantries for moving the object. Spatial localization in three dimensions is achieved by application of three orthogonal time dependent linear gradient magnetic fields in the presence of a continuous string of closely spaced phase alternated RF pulses, being of a short duration, in the order of 20 microseconds. The received signal is a combination of MR signals of nuclei around the intersection point of three gradient fields and the frequency modulated signals of the surrounding regions.

The RF pulse scheme, known as 'steady-state free precession (SFP) provides a continuos and large component of the transverse magnetization which contains

nent of the transverse magnetization which contains both the time dependent and time independent MR signal. The time dependent signal is removed by a low pass filter.

MR point scanning methods are no longer used for imaging in-vivo objects because of the very long scan times. They are only applied in cases where spectral MR information with high resolution is desired such as for the topological examination of the phosphoric structure or measurements of isotops which emit a very weak MR signal.

One advantage of sequential point methods is their simplicity of data processing which unlike the other methods, often enables them to be performed without the necessity of a computer.

4.2.2. Sequential Line Methods

There are two methods classed as sequential line methods. In both these methods the recorded MR signal is restricted to the response from a line of n elements in the object. Fourier transformation of the MR response to suitably applied radiofrequency pulses gives the MR pulses from the n voxels in the line at the same time. The line may then be traversed through a defined slice of the object, and the image of that slice recorded. If the other instrumental parameters are kept the same, sequential line methods will be n times faster then sequential point methods.

One of the sequential line methods is the sensitive line or multiple sensitive point technique which was developed by Hinshaw as a more efficient version of the sensitive point methode. It utilizes two orthogonal time dependent magnetic field gradients, an SFP pulse sequence, and signal averaging to spatially localize the MR spectrometer sensitivity to a line as in the sensitive point method. The distribution of spins along the line is determined by applying a third linear magnetic field gradient in the direction of the line that is time independent. The effect of the static gradient is to vary the Larmor frequency of the spins linearly with position along the line. Thus the MR frequency spectrum directly corresponds to the spin density distribution along the line and can be obtained by Fourier transformation of the time averaged FID of the MR signal.

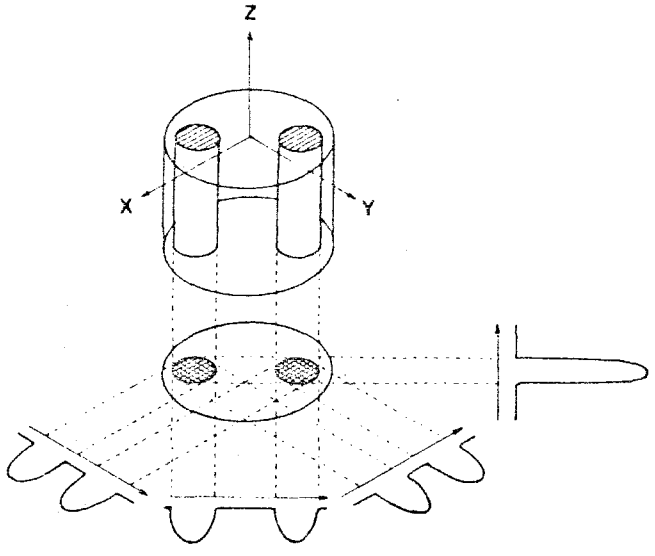
This method is very time consuming during data acquisition, so that it is nc more used in MR imaging procedures.

4.2.3. Sequential Plane Imaging

The Sequential plane methods provide a good compromise of image quality with short acquisition time. They also allow multiple slices to be reconstructed without any increase in data acquisition time or loss of image quality. With these methods any of three planes, transaxial, sagittal or coronal can be selected.

There are five sequential plane methods which can be used in MRI. One of these methods makes possible to sort the MR signals of all the excited nuclei spins in a plane at the same time. By mean of this method the two dimensional spatial variation or image of a physical property of an object can be reconstructed from a series of one dimensional projections of the parameter that are recorded at different orientations relative to the sample. This principle forms the basis of highly succesful X-ray CT imaging method. It is applied by "Lauterbur" to MR and called as "2D Projection Reconstruction Method". However, in CT technique real X-ray projections are used, whereas for MR imaging projections of the samples' should be produced artificially. A one dimensional projection can be obtained by recording the MR spectrum in the presence of a linear magnetic field gradient as shown in Fig.4.3.(35)

Fig.4.3. Projection Reconstruction Imaging



Multiple projections are obtained by changing the relative orientation of the gradient. The data are then manipulated in a computer by using a standard projection reconstruction algorithm to give an image resolved in two dimensions. The use of pulsed Fourier Transform enables the simultaneous recording of data from the entire plane for each projection, as in sequential line methods. The electronic reorientation of the gradient by feeding the voltages corresponding to the vector components of the gradient into the x- and y- gradient coils is an important improvement over other methods. The production of an image with this method is illustrated in Fig.4.3.

Fig.4.4. Spatial imaging with 2D Projection Reconstruction Method

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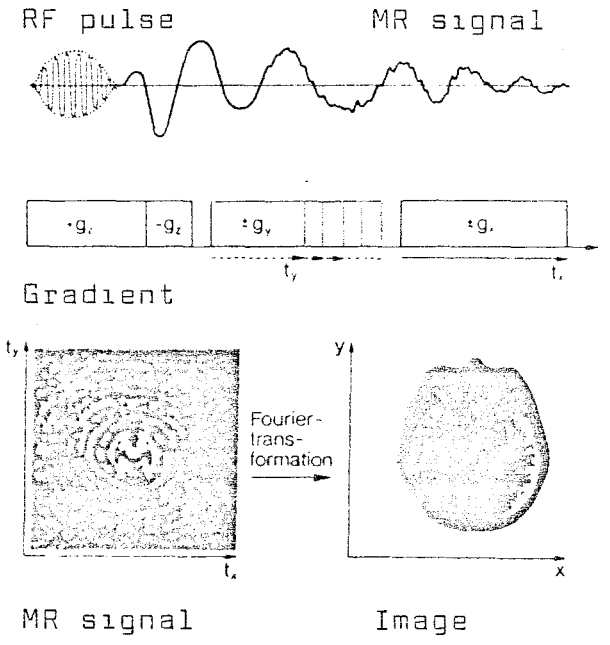
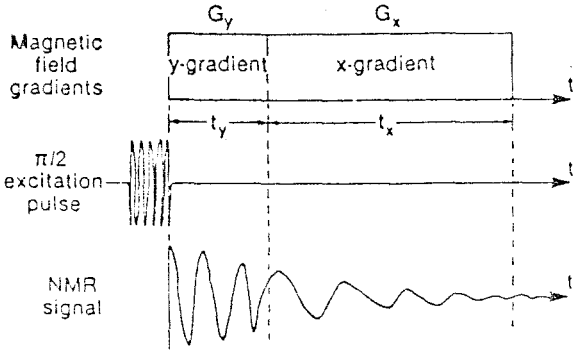


Another method for reconstructing an image in two dimensions is the "2D(two dimensional) MR Fourier Imaging" devised by Kumar. The technique utilizes a sequence of switched magnetic field gradients applied during the FID, combined with 2D Fourier Transform Method. The imaging scheme is illustated in Fig.4.4.

Fig.4.5. Principles of 2D Fourier Imaging Method



This method is based on the fact that in an MR experiment not only the intensity, but also the phase of the measurement is accessible. So, local information for any direction in space can be recorded in the form of a phase angle in the MR signal. The measured values, accumulated by Fourier Imaging Technique represent the lines of a hologram as shown in Fig.4.5. The complete holcgram is built up from many individual measurements which follow each other in sequence of time intervals. A two dimensional Fourier Transform serves for image reconstruction.

As shown in Fig.4.5, spins in the sample are excited at time t=0 by a $\pi/2$ RF pulse. Orthogonal linear gradients gx and gy are then successively applied for durations tx and ty, with the FID being recorded in the interval tx.In the subsequent experiments, the period ty is systematically varied ny times to collect ny FID's represented by S(tx,ty). The two dimensional signal function S(tx,ty) contains all of the information necessary to reconstruct a two dimensional image. T1 images can be produced by adding a pulse suitably spaced at the beginning of each experiment. A characteristic of this imaging method is that the signals are acquired only during a part of the FID, which results in some reduction in sensitivity relative to methods such as projection reconstruction that observe the entire FID. The use of aquadratic or higher order z gradient has been suggested for localization in the third dimension.

In the third sequential plane method,"planar imaging" thin strips in the defined plane are excited and read in a carefully oriented oblique field gradient.Because the strips are narrow compared with the spaces between them, much potential MR signal is wasted. The method has therefore been superseded by the fourth technique of this group, namely the "2D echo planar method". Here, an alternating gradient generates periodic echoes whose Fourier transform gives an MR response that behaves as if the the resonant nuclei were heaped in strips across the defined plane, without any waste of MR signal. The signal may then be read in an oriented static gradient to give the image from the entire plane from one FID.

The final sequential plane method,"rotating frame imaging" has not been fully developed at present.

4.2.4. Three Dimensional Imaging(Simultaneous Methods)

The MR experiment is inherently sensitive to whole volume measurements, including point, line, and planar imaging where considerable effort is required to restrict the spectrometer sensitivity to lesser regions of interest. In three dimensional imaging methods which are extensions to three dimension of sequential plane methods, the subsequent decoding and reconstruction of the three dimensional spatial information, with an nz fold increase in data over planar methods, poses a formideable computational problem if high spatial resolution is to be maintained. For this reason, only one such method, namely, "3D (three dimensional) projection reconstruction method" has actually been demonstrated.

Three Dimensional Projection Reconstruction Method is the obvious extension of the two dimensional projection reconstruction procedure. Instead of a plane localization, the gradient is reoriented in all three dimensions. The one dimensional projections obtained for each gradient orientation contain signal components from the entire sample. The three dimensional image is reconstructed using a three dimensional version of the reconstruction algorithm, and displayed as a series of planes in any desired orientation.

3D Fourier Imaging, Multiplanar Imaging and 3D Echo Planar Imaging are the other simultaneous methods.

Three dimensional Volume Imaging currently has two problems which limit its use in clinical imaging. The data acquisition time required to obtain the necessary signals from a large volume in order to produce an image of equal quality to Sequential Plane, is long. Further, reconstruction time is excessive. The time between starting the measurement and viewing the first image is 1 - 1.5 hours.

On the other hand these methods have the merit of simplicity, require no complicated computations and no computer.

4.3. Measurement Techniques

There are several measurement techniques which constitute a base for the MR imaging. According to the factors that effect the image quality(e.g. the tissue that is scanned), one of these techniques is chosen.The mostly used methods are explained in the following sections.

4.3.1. Free Induction Decay (FID)

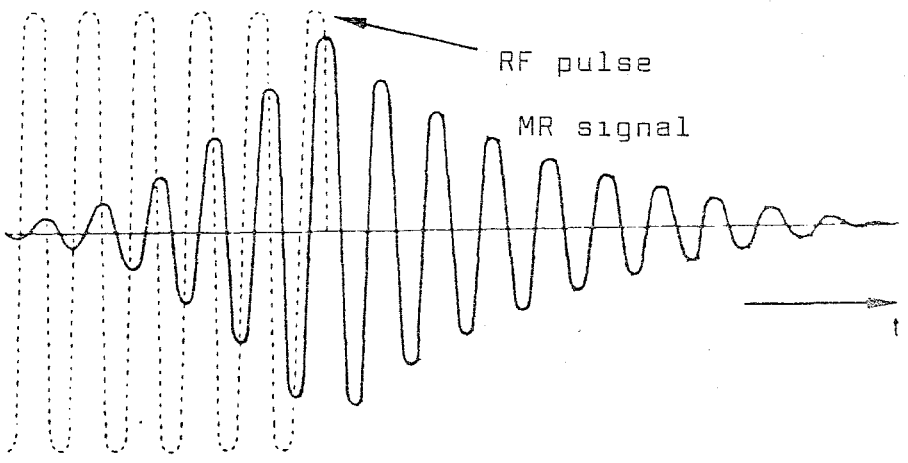
FID is a simple MR imaging technique where a brief pulse of RF, e.g. a 90^{0} pulse, at the resonant frequency rotates the M vector out of the z-direction. When the pulse is terminated one observes an

oscillating sin-wave signal ($\sin(\omega_c t)$) which decreases in amplitude in an exponential fashion with time ($e^{-1/T2}$) These results are referred to as the free induction decay.

The first FID is relatively strong because the protons were aligned with B_o. In order to get another projection in x-y plane, the same 90^o and z-gradient pulses have been repeated.

As seen in Fig.4.6 after the RF pulse, a transient MR signal results that will decay towards zero with a characteristic time constant T2. That means, FID is a method for measuring the spin-spin relaxation time T2.

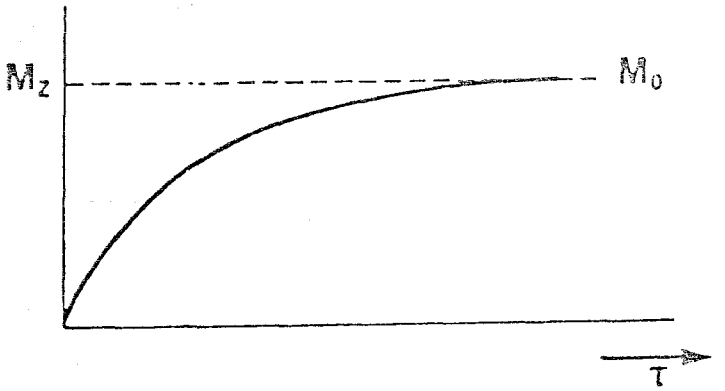
Fig.4.6. Free Induction Decay



4.3.2. Inversion Recovery (IR)

Inversion Recovery is a pulse MR technique which can be incorporated into MR imaging, wherein the magnetization is inverted at a time on the order of T1 before the regular imaging pulse-gradient sequences. In this method a 180^{0} RF pulse is applied to invert the magnetization to the -z direction which then starts to decay back to the positive z direction with time constant of T1. No signal can, however, be detected unless the magnetization has a component in the xy-plane. A certain time lap after the 180 pulse, a 90^{0} pulse is applied which rotates the remaining magnetization onto this plane. The received signal is then dependent on T1 and with careful timing, it can be strongly dependent on it. The characteristics of IR can be seen in Fig.4.7.

Fig.4.7 Following a 180^{0} pulse the net magnetization will return to the equilibrium value M_0 with an exponential decay determined by the longitudinal decay constant 1/T



Here, it is assumed that the pulse repetition time between two successive measurements is long relative to the longitudinal relaxation time T1. Again, two or more data sets obtained with different interpulse delays can be combined mathematically to form T1 or spin density maps. The inversion recovery pulse sequence highlights T1 variations in the sample more strongly than the saturation recovery technique does but the price paid is a longer scanning time (or a lower spatial resolution), since a delay time of at least 3 T1 should be allowed before repeating the 180° - 90° pulse pair if errors in T1 values are to be avoided.

It is important to realize that for most biological samples, relaxation times are characterized by multiexponential decays. Care must therefore be exercised in interpreting relaxation data obtained from T1 and T2 maps, as the results are usually weighted averages of a family of values, which depend on many different cellular microenvironments in which the nuclei find themselves.

4.3.3. Saturation Recovery

The Saturation Recovery technique usually consists of applying a series of equally spaced 90° pulses. The pulse interval T is of the order of the average T1 value of the sample.

In this method,the magnetization is saturated i.e. the alignment of the nuclear moments is disturbed to such a degree that no magnetization can be measured externally. After the interval T, in which the magnetization builds up again with time constant T1, the excitation of the nuclei takes place. After a further interval the reading of the signal begins.

The signal S for the image is derived from the following equation:

$$S = \rho e^{-\tau/T2} (1 - e^{-t/T1})$$

where ρ : proton density
τ : echo time

If two or more images are obtained using different values of T, both a T1 map and a spin density map uncontaminated by relaxation effects can be generated from the several data sets.(87)

By using this method, relaxation times T1 or T2 are displayed instead of the direct display of the magnitude S as an image. Thereby a fundamental distinction is made between the display of the MR signal S and calculated images, which directly reproduce the relaxation times.

4.3.4. Spin Echo Sequence

Spin echo sequence is the reapperance of an MR signal after the FID has died away, as a result of the effective reversal of the dephasing of the spins by such techniques as reversal of a gradiant magnetic field. Multiple spin echoes or a series of spin echoes at different times can be used to determine T2 without contamination by effects of the inhomogeneity of the magnetic field. For this purpose an initial 90° RF pulse will be given which moves the magnetization into the xy-plane where it decays with a time constant T2. Unfortunately magnetic field inhomogeniety disturbs this process but it is found that if a further 180° pulse is applied to flip the magnetization, still in the xy-plane, to the -x direction, the effect of the field inhomogeniety cancels out leaving only a T2 dependence. Again a signal which is largely dependent on T2 can be obtained.

4.3.5. Chemical Shifting

The Larmor Frequency equation indicates that all the nuclei of a particular element in a homogeneous magnetic field will have the same resonant frequency. A frequency spectrum of the MR signal should therefore displlay a single narrow peak. For the nuclei of very simple liquids (e.g. protons in pure water) this is indeed so. For more chemically complex samples, however the applied magnetic field is subtly altered around

some nuclei as a result of "shielding" currents which are associated with the distribution of electrons around adjacent atoms. Such alterations cause shifts in the resonance frequency ("chemical shifts") and contain valuable information concerning the molecular structure in which the nuclei exist. Although these frequencie shifts are sufficiently small that they are not generally detectec or employed in current MR image formation, they are very valuable in chemical studies and permit in vivo investigations of biochemical processes using external detectors such as surface coils.

4.4. MR Spectroscopy

At this time, spectroscopy as a useful method in clinical MR imaging is still in an early investigative stage. However, it may become important in the near future.

Not all protons have exactly the same resonant or Larmor Frequency. The proton resonant frequency varies as a function of its chemical environment depending upon the physiological and psicological state of the human being. MR methods, which are sufficiently sophisticated to detect MR signals from protons with different resonant frequencies,can be referred to as " spectroscopy ".

In spectroscopic techniques, the individual free induction decays are typically detected and then

transformed into absorption peaks or a spectrum via Fourier transform methods. The ability to detect seperately or resolve protons with varying resonant frequencies also allows for the ability to obtain their individual T1's, T2's, etc.

In radiological studies, the most important signal differentiation is that between water and fat protons. If one could differentiate between water and fat protons, it would then be possible to characterize directly the fat content of tissues. This could be of great utility in imaging methods. Spectroscopy will also play a role if radiology begins to employ methods for imaging nuclei other than protons.

4.5. MR Images Other Than H-Imaging

All of the commercial MR units that are produced for medical diagnostic purposes use hydrogen imaging. The MR units with phosphorus, sodium or fluorine imaging principles are not yet totally developed, and are mostly used in laboratories for researche purposes.

4.5.1. Phosphorus Imaging

One reason for the interest in phosphorus is that this element is sure to provide important details about the body's biochemistry, details not available with other imaging methods. Adenosine triphosphate (ATP), for example, is a rich energy source for the body's cells. Metabolism, the breaking down of nutrients for

cell construction or maintanence, is fueled by the breakage of phosphorus' chemical attachments. MR may one day be able to detect abnormal distributions of phosphorus to aid in diagnosing any of a number of biological malfunctions. Phosphorus imaging promises to provide new insights into how human tissue functions.(76)

Phosphorus imaging may also be useful in cancer studies. Physicians now have no way of quickly determining the growth rates of certain brain tumors.It is reasonable to expect that fast growing tumors would have large ATP reserves. Besides aiding diagnosis, such imaging can be performed during therapy so that physicians can tell whether the treatment slowsdown the growth of the tumor by simply comparing phosphorus concentrations over a period of time. Likewise, phosphorus imaging may aid cardiologists in spotting weakened hard muscle tissue, urologists in predicting the success of kidney transplants, and pediatricians in determining if high risk new borns are suffering irreversible brain damage due to oxygen deficiencies.

The increased use of ^{31}P MR in biological research in the past years owes much to magnet development. In the continuing search for greater sensitivity, the MR spectrometer designer has usually opted for higher magnetic field strengths, stretching superconducting magnet technology to the limit.

4.5.2. Sodium Imaging

Sodium imaging may aid in the treatment of strokes. Sodium concentrations increase dramatically at stroke sites, a phenomenan thought to stem from fluid accumulation around the damaged tissues.

As the sodium level rises, so does the intensity of the MR image. Such changes may provide not only a new diagnostic tool but also a means of determining how stroke sites are responding to treatment.

As an example, images of the head show the distribution of the free sodium. While intracellular sodium of the brain practically provides no signal, a high signal intensity is found in the inner and outer subarachnoid spaces and in the hyaloid body of the eye.

4.5.3. Fluorine Imaging

Fluorine occurs in the human body only in relatively small amounts. But this scarity is a good advantage: It means that under certain conditions, fluorine introduced into body will produce strong signals unobscured by background "noise".

Many important surgical anesthetics contain fluorine. But how these agents effect the human nervous system and induce deep sleep remains unknown. Little is known of where in the brain the anestetic goes or how long it stays there.

Certain fluorine-containing chemicals, such as 5-fluoroacil, appear to be useful in threating and destroying tumors. MR could thus help visualize sites where the drug is or is not at work.

Blood flow studies in deep, relatively inaccessible organs, such as the liver might also be realized with recently developed blood substitutes. These compounds, based on biologically inert fluorinated hydrocarbones, are easily distinguished from surrounding tissues. Therefore they may one day be used to determine flow rates within smaller vessels deep inside the body.

5. CONSTRUCTION OF AN MR SYSTEM

The basic instrumentation needed in an MR system is as given below:

a-) Magnet,

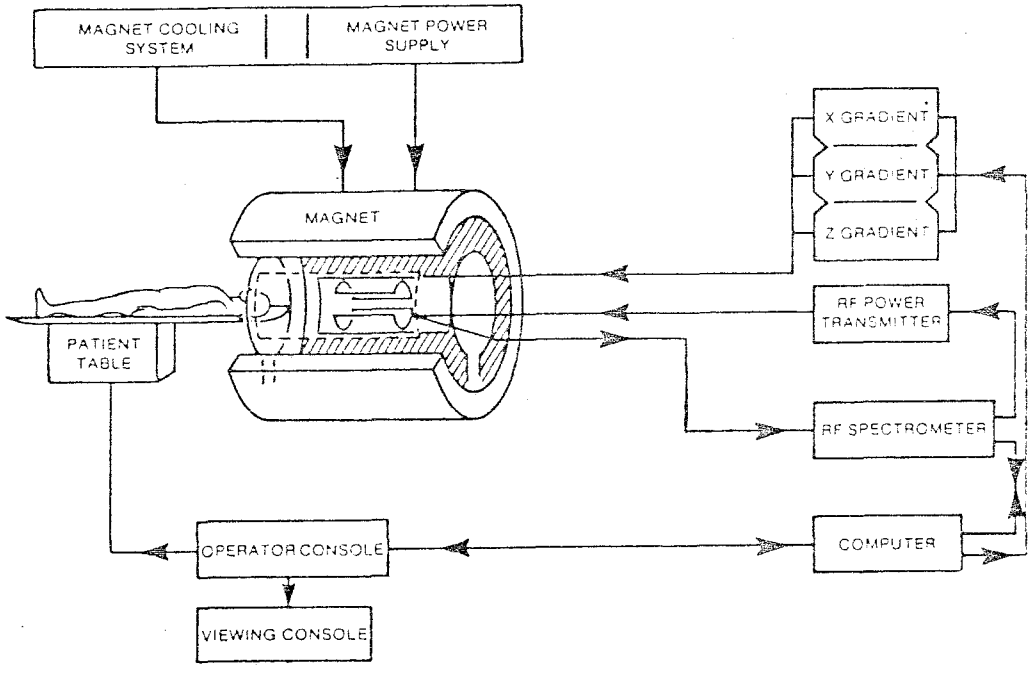
b-) Gradient System,

c-) RF Coil

d-) Computer

A schematic diagram of such an MR system is shown in Fig.5.1. Each of the above mentioned parts will be explained in the following sections seperately.

Fig.5.1. MR Imaging schematic



5.1. The Magnet

The basic element of an MR imaging system is the magnet which generates the static field Bo. To obtain images of the human body, the magnet must have a large enough bore to encase the body and the coils that surround the body. For the whole body systems three choices exist:

a-) Permanent magnet,
b-) Resistive electromagnets,
c-) Superconducting magnets.

The decision is one of economics and the desired field strength. Below these three kinds of magnets and their advantages and disadvantages will be discussed.

5.1.1. Permanent Magnets

A great number of alloys show ferromagnetic characteristics. A magnet which is produced out of such a material has the advantage of not having an external generator. Thus, the cooling system also becomes redundant. (41)

For construction, units of permenently magnetized

material are stacked together to act as the source of the field. These units will be relatively small compared to the total volume of the complete magnet and will be accurately machined and assembled. Each unit will have N and S poles at its ends and the object of the assembly will be to give a finished magnet with N and S poles on opposite sides of an air gap of a whole body size.

The field strength in the gap of such magnets is inversely proportional to the width of the gap, but it can be increased by routing the field from more source material through the gap.

The disadvantage of these kind of magnets is their huge weight. They weigh around 110 tons. Such weight is very hard to transport and install in the MR device.

The volume of material in such a magnet is about $12m^3$ and stands on an area of floor of about $6m^2$ giving a loading of 16 tonnes/m^2. These values of loading factor and total load cannot normally be supported by elevated floors in most buildings and so the magnet must be ground based on a suitable floor.

The main virtues of permanent magnets are the low running costs and the lack of peripheral magnetic field problems. It has to be stressed however that, for presently available materials that can be used in a commercial scale, the maximum field strength that can be achieved is 0.1-0.3 T, which appears as a great disadvantage.

5.1.2. Resistive Magnets

Resistive magnets fall into two broad classes. The first is purely resistive and consists of circular coils of wire rigidly held into a support framework. This framework ensures the correct placement of the coils necessary to maintain the homogeneity of the magnetic field. The second class of resistive magnets has also circular coils, but in addition they have iron or steel outside the windings. The iron is in the return path of the magnetic field outside the windings and is magnetized by this field. The magnetization then produces a magnetic field which augments the field at the center of the system.

It is during the design of the magnet that the exact placings of the field sources(coils and irons) are determined. The criteria used in the design are the size of the required homogeneous volume and the working access bore within the windings. The design should then be optimized to work at minimum power consumption.

The initial assembly of the magnet and the subsequent re-assembly at the operating site can be done to a great accuracy. This is, however, not enough to generate a field of the required homogeneity. The reasons are that, small and unavoidable irregularities occur, and most buildings where the MR devices are installed contain unpredictable quantities of steel. Both of these facts produce distortions of the magnetic field making the system unfit for the designed application.

There are some disadvantages of resistive magnets that make them not very suitable for medical devices. Electrical power requirements and water cooling difficulties are two of these.

A typical 0.15 T purely resistive whole-body magnet weighs approximately 2 tonnes, and requires in the order of 60kW of continuous electrical power. The water cooling system, has to fulfill two conditions: no surface of the magnet should rise above 40 C, and the temperature should be stable to better than 0.5 C. This latter condition implies that there is a warm-up time for the magnet. In practice, some magnets are ready to operate after 20 minutes of warm-up, where for some others it takes many hours. The long warm-up times are not acceptable for a routine instrument thus the necessary temperature must be maintained 24 hours per day. Three methods are used for this purpose, one is simply to run the system continuously, the second is to run the system at low power with the circulating water, and the third is to circulate warm water from a low cost source. The warm-up time is obviously of considerable importance to the running costs.

To remove 60kW power with a reasonable control of temperature requires approximately 60lt./min of cooling water. The inlet temperature of water must be constant and the exhaust water must be disposed of either by throwing away or recirculating through a cooling system. This can only be obtained by a closed circuit water system which rejects heat to a water stream which will eventually be discarded.

An important disadvantage of resistive magnets is their low magnetic field strength. The maximum value for such a magnet is 0.2 T, a value which is not sufficient for some of the in-vivo examinations.

5.1.3. Superconducting Magnets

If certain alloys are cooled down at a temperature near the absolute zero degree(4° K = -269° C), they show a very small electrical resistance ; they become " superconducting ". High currents can flow through a coil of such a material without being dissipated. Magnets produced out of this kind of materials are called superconducting magnets. (24)

The greatest advantage of a superconducting magnet is, that it doesn't consume any energy in order to conserve the magnetic field. The only necessary condition is to keep the temperature at 4° K, which is achieved by using liquid helium.

A superconducting magnet will be 'loaded' to its working magnetic field once, and then the field always keeps this initial value.

Today, there are two kinds of materials which are used for magnet construction. One group consists of conductors on the "Niob-Titan" basis. These become superconducting below 9° K. The second group are the conductors on "Niob-Tin" basis. With Nb_3Sn which must be kept at 4.2 $^{\circ}$K, magnets up to 15 T can be produced.

The magnetic field in a superconducting magnet is maintained by large currents circulating in the windings. This represents a stored energy of 0.5 - 5MJ, changing according to the field strengths. If the current is interrupted in any way or the magnet warms above a certain limit, then this energy will be dumped within a few seconds to the liquid helium. This has quite exciting results and the whole effect is called a ' magnet quench '.

Superconducting magnets represent the only way to provide field strengths in excess of 0.3 T for the whole body systems. In the lower magnetic field range, it can be discussed whether resistive, permanent or superconducting magnets are to be chosen.

The magnet itself is a complete superconducting loop with two main features. The most obvious of these is the magnet winding. The other is known as the superconducting switch. This is a section of the superconductor surrounded by a heating element and embedded inside a thermally insulating jacket. When the heater is on, the superconductor loses its superconducting properties and becomes a piece of normal metal with a fairly high resistance(50-100ohm).

The whole sequnce of initially energizing the magnet takes between 3 and 12 hours. The peak power consumption during this period should not exceed 5kW.

The decay rate of the magnet in practice is likely to be between 1×10^{-8} and 6×10^{-8} T /hour. If the temperature is kept on the ideal values, the current flows approximately 1000 years and the magnetic field also remains nearly at the initial strength.

Liquid nitrogen consumption will be of the order of 200-250 lt/week and can be obtained from a bulk storage system either directly by a fixed, insulated transfer pipe or by a transport dewar of suitable capacity. Liquid helium has very small latent heat of vaporization. The typical consumption of liquid helium is between 50-100 lt/week, while the capacity of the cryostat is 200-300 lt depending upon the design. This gives a refill period of between 2-6 weeks.

5.1.3.1. Magnetic Hazards

Since the superconducting magnet doesnt't have a power supply connected under normal operating conditions, the field cannot be reduced in the case of a magnetic hazard or an emergency by this means. The method of emergency magnet discharge is either to open the superconducting switch which will reduce the field over 5-20 minutes or to heat a section of the magnet windings, thus quenching the magnet within a few seconds.

The cryogens are normally under vacuum except for the helium and nitrogen cans. In cases of failure of the internal cans, the vacuum vessel must be protected against overpressure and similarly the helium and

nitrogen cans must also be protected against overpressure. The latter is done by 'bursting discs' which release any excess pressure to the atmosphere. Since this could involve large quantities of gas, the magnet room should be well ventilated.

5.1.4. The Optimum Field Strength of Magnets

In the short history of MR, there has been a great debate about the optimum field strength of the magnets for desired image quality. However,there is no one field strength that is optimal for every situation. Before making a decision about the field of the magnet, the projected clinical uses of the system and the practical concerns of the institution must be considered. (86)
The considerations are :

a-) Applications,
b-) Image quality,
c-) Throughput,
d-) Capital and site-related costs.

5.1.4.1. Applications

The current clinical application of MR is proton imaging. It is generally agreed that spectroscopy of other nuclei as a routine clinical procedure will not occur for several years. Today, obtaining a system that will do spectroscopy makes more sense for research institutions than for hospitals.

5.1.4.2. Image Quality

The image C/N (contrast to noise ratio) beyond 1.0 T depends only slightly on field strength. Most, if not all, of the gains that can be practically realized occure at this field strength.

The S/N gains of even higher fields are much more pronounced for spectroscopic applications using rare nuclei like phosphorus and sodium. Also, the increase in chemical shift at high fields facilitates spectroscopic investigations. Therefore, 2.0 T is the best available field for this type of research.

5.1.4.3. Throughput

Once sufficient S/N has been realized to allow single-acquisition imaging, there will be not great field-dependent improvement in patient throughput. Because of the minor differences in C/N beyond 1.0 T, systems operating at this field strength or higher won't likely exhibit any benefit on throughput caused by the field strength itself.

5.1.4.4. Capital and Site-related Costs

Higher field strength systems, in general, are more costly to obtain and to install than those operating at lower fields. This extra cost may be worthwile if it results in tangible clinical benefits.

5.1.4.5. Summary

The optimal field strength changes from case to case according to the needs of the medical institution. 1.0 T may well be the optimum field strength for a hospital where MR is used for routine clinical diagnosis only(85). It makes good use of the gains derived from increasing field strength and offers sophisticated image quality with clinically realistic cost and siting requirements.

Multinuclear spectroscopic investigations are best performed at the highest field strength possible. Therefore, a 2.0 T system should be the choice for this application.

5.2. Gradient Fields System

Gradient fields are normally weak fields up to +/- 15 gauss. However, they can be changed very quickly. Gradient fields are superimposed on a strong static field so that spatial coding is achieved within

a large volume; in simple terms, to differentiate one location from another during imaging.

The gradient system consists of a set of three orthogonal DC coils, generating the three principal linear gradients:

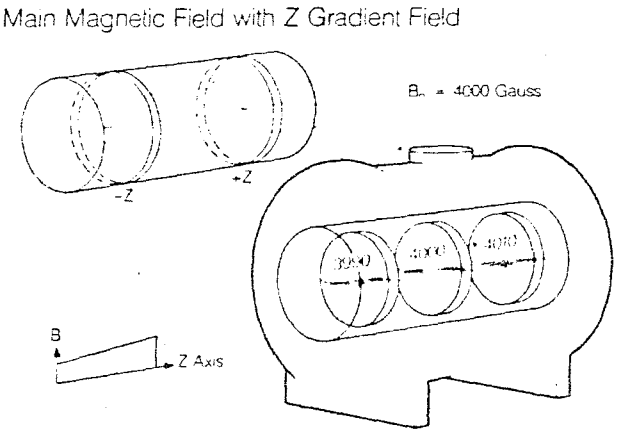
$$\partial Bx/\partial x = Gx \quad ; \quad \partial By/\partial y = Gy \quad ; \quad \partial Bz/\partial z = Gz$$

Above mentioned gradients, as they are required for projection-reconstruction techniques are implemented by suitably combining two or all three gradients. The gradient coils are driven by their own power supplies and can be switched under computer control. Both linearity and stability are critical image quality parameters.

We can discuss the effects of the gradient fields by considering one of them,e.g. z-gradient(see Fig.5.3)

Two selenoid gradient coils are wound on the gradient coil set. One coil is wound in the same direction as the main winding. The other coil is wound in the other direction. When a current is passed through the two coils, two opposing magnetic fields are produced. These are shown in the top left of the figure.

Fig. 5.3. Effects of the z-gradient field (85)



In the magnet these two fields are superimposed upon the main static field. The result is shown to the bottom right of the figure. The +Z-gradient coil increases the field strength to a maximum, here shown as 4010 gauss. The two fields were in the same direction and have simple been added. In the centre of the magnet the two gradient fields cancel and accordingly the field here remains unchanged at 4000 gauss, the original static field. As we move down to the lower end of the magnet where the -Z gradient coil is located, the -Z gradient cancels with a part of the main field. The resultant field is then 3990 gauss.

In every case the flux still travels in the same direction down the bore of the magnet. If we were to measure the field along the Z axis we would see that a linear gradient existed as shown in the bottom left of the figure. At any point on the Z axis the field is

constant within a thin disk area of the bore.

Exactly the same consideration can be made for X and Y-gradients. Therefore it is not explained seperately for those gradients, but the figures which are helpful for the consideration are given below.

Fig. 5.4 Effects of the X-gradient

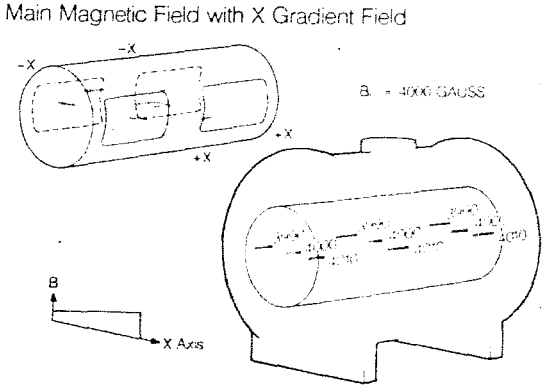
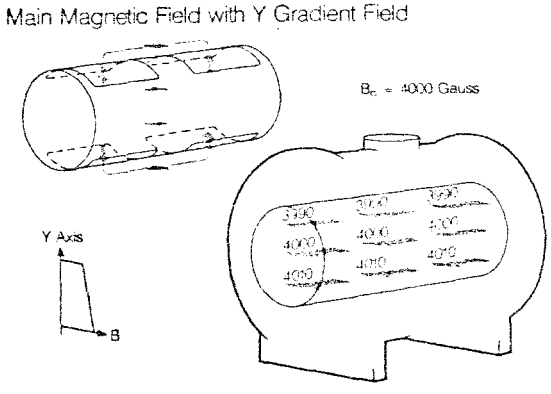


Fig. 5.5. Effects of the Y-gradient

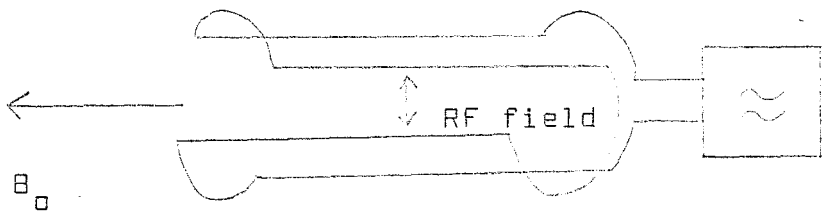


5.3. RF Coils

The purpose of the RF coils is to generate the B1 field requisite for the excitation of the magnetization and for picking up the free induction decay signals. As most magnets provide a radially symmetrical field, accommodating the patient with his axis centered along the direction of the magnetic field, a pair of saddle shaped RF coils is mounted on a coaxial tube in such a manner that B1 acts perpendicularly to the main magnetic field. Ideal RF coils provide a B1 field amplitude that is constant across the imaging volume, therefore uniformly exciting the nuclei in the specimen.

Transmitter is the section of the RF system that produces the highly stable radio frequency and amplifies it to a level suitable for pulsed excitation. There is also a receiver part in the system where the signals coming from the patients body are detected. For body examinations one common coil is used for both transmission and reception. A small receiver coil is used for brain studies in order to improve the S/N. The pair of saddle coils antenna is shown in Fig. 5.6.

Fig. 5.6 Pair of Saddle Coil Antenna



5.4. Computer System

The computer system of an MT is generally subdivided into four sections as shown:

a-) System Control,
b-) Data Measurement Unit,
c-) Image Processor,
d-) Diagnoctic Consoles.

System control section acts as the system manager. The host computer with associated software does not only supervise the entire system but also performs a large number of further functions such as system operation, software and image data storage, image evaluation and image manipulation.

Data Measurement unit provides all functions which are necessary to obtain an MR signal, to convert it to a digital value and to transfer the data to the image processor.

Image processor receives digital data from the RF system within the data measurement section, performs corrections such as offset and reconstructs the image after that the raw data have been stored on disk.

Diagnostic consoles are the units where the operator can manipulate image reconstruction, image archiving, image manipulation and other necessary processes.

5.5. Available MR Systems in the Market

Today, there are more than 15 companies on the world that produce MR systems. These systems have some differences in the type of magnet, in the construction, and in the operating modes. A table showing a comparison between MR systems of some manufacturers is given below.

Table 5.1. Comparison between different systems

Manufacturer	Type of magnet	Field strength	Spectroscopy	Operation mode	Slice
G. Electric	Supercond.	0.3,0.5, 1.0,1.5T	Yes	IR,SER 2D Four.	?
Technicare	Supercond. Resistive	0.5,1.5 0.15	No	IR,SER 2D,3D Echopulse	10-25
Fonar	Permanent	0.3	No	IR,SER,2D	8
Elscint	Supercond.	0.35,0.5 1.5	No	IR,SER	5,10, 20
Siemens	Supercond.	0.5,1.0, 1.5,2.0	Yes	IR,SER 2D,3D Multislice	5,10, 15,20
Picker	Res./Super.	0.15,1.5	Yes	IR,SER,2D	5-20

6. MEDICAL APPLICATIONS OF MAGNETIC RESONANCE IMAGING

The ability of Magnetic Resonance Systems to obtain direct transaxial, sagittal, and coronal images besides the multiplicity of techniques of data acquisition makes possible the demonstration of any abnormality in the human body which cannot be shown by any other imaging modality. With MRI, it is possible to display soft tissues in many different ways. By selecting either spin echo or inversion recovery techniques and varying the parameters of the pulse sequence different characteristics of the tissue can be emphasized.

Information derived from MR in most cases is currently comparable to that obtained by other diagnostic imaging methods, including Conventional X-ray, Computed Tomography, Ultrasonography, and Nuclear Imaging Systems.

6.1. Comparison Between MRI and Other Important Diagnostic Systems

In the following sections four different diagnostic imaging techniques will be discussed. A comparison of each of these imaging modalities with Magnetic Resonance will be given.

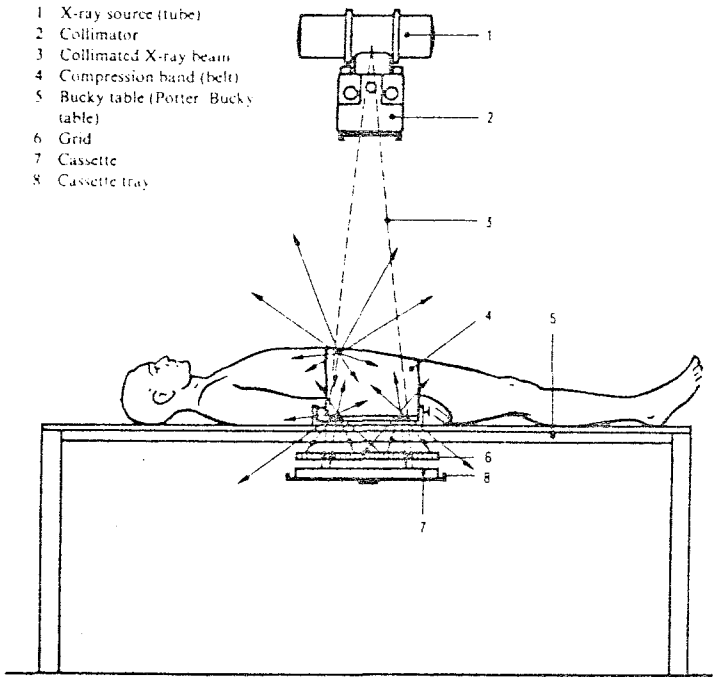
6.1.1. Conventional X-ray Diagnostic Systems

X-rays are waves with a length of 10^{-8} cm. They pass through matter and attenuate thereby with a certain degree, depending on the atomic structure and density of the matter, where dense bony substances absorbe more radiation than soft tissues. The passed radiation falls on a special X-ray film and causes different degrees of blackening.

The only parameter in the conventional radiography is the density of tissues. With this technique, only a two-dimensional image can be obtained from a three dimensional object.A great advantage of X-ray systems are the practical use and the short duration of examinations. These systems have nearly no operation costs. A simple X-ray device is shown in Fig.6.1.(53)

Conventional X-ray devices are the mostly used diagnostic imaging systems in medicine. But they are in no way comparable with MRI, or even with CT. The most important reason for this is that X-ray films can be taken only from one plane, and that they are very poor in contrast detail with respect to MR and CT.

Fig.6.1. X-ray equipment consisting of a bucky table and an X-ray tube



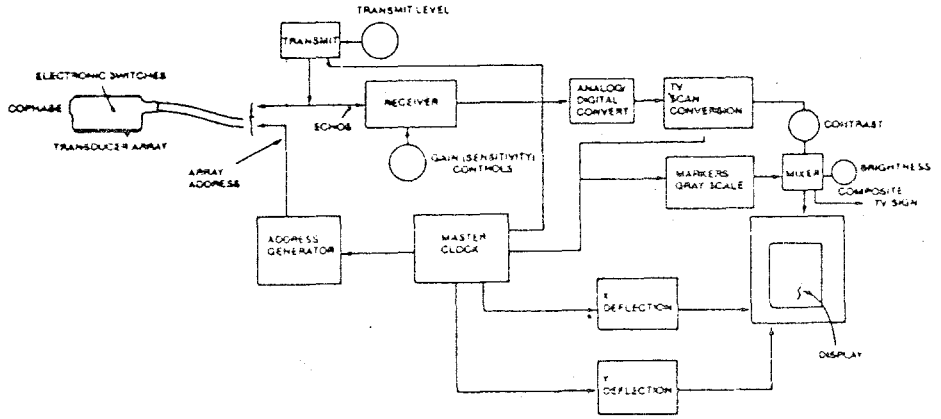
6.1.2. Ultrasonography

Ultrasonography is the term applied to mechanical pressure waves with frequencies above 20 kHz. The ultrasound beam behaves similarly to a light beam in that it can be reflected, refracted and diffracted. It can also be absorbed by the media it traverses. In this process the kinetic energy of the beam is converted to heat, resulting in a decrease in the intensity of the ultrasound energy.

The part of the ultrasound beam which is reflected as an echo provides a rich source of diagnostic information. These reflections occur at interfaces between materials having different acoustic properties. The echoes are analyzed first with regard to their site of origin and second with regard to their intensity. It can be said that ultrasonography is a technique for recognizing the boundaries of objects rather then themselves.

A diagram explaining the principles of ultrasonographic devices is given in Fig.6.2.

Fig.6.2. Block diagram of an ultrasonographic device.



Ultrasonographic examinations are done easily and they don't take a long time. Furthermore the acquisition and operational costs of these instruments are relatively low. However, for many organs ultrasonographic images are not enough for making a detailed diagnosis.

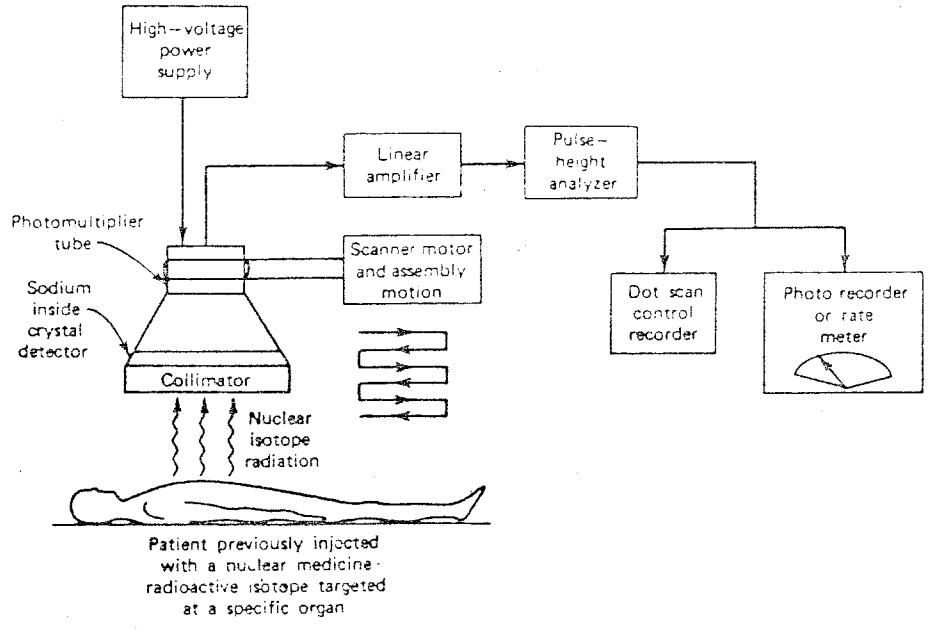
6.1.3. Nuclear Medicine

The function of Nuclear Medicine is to prove the existence of radioactive indicators in vivo and in vitro in order to receive information about the transport-, distribution-, metabolism-, and excretion-processes of the human body.

For this purpose it is necessary to measure the emitted beam with a high precision, which shows the presence of the indicator.

Until some years ago, rectilinear scanners were mostly used in nuclear medicine. Today gamma cameras and PET scanners(Positron Emission Tomography) have taken their place. Gamma cameras are applied for routine examinations where PET is mostly used for research purposes. The simple operation principles of a rectilinear scanner is shown in Fig.6.3.

Fig.6.3. Operation principles of a rectilinear scanner



In nuclear medicine indicators concentrate in specific organs, which brings the advantage of localizing the examined organ.

Nuclear medicine techniques provide a fairly improved contrast and a sophisticated contrast media sensitivity compared to radiological techniques. However, they suffer from poor spatial resolution(0.5cm). The fact that nuclear medicine can show physiology as opposed to anatomy is an important advantage. In that respect it can be compared with MRI.

6.1.4. Computed Tomography (CT)

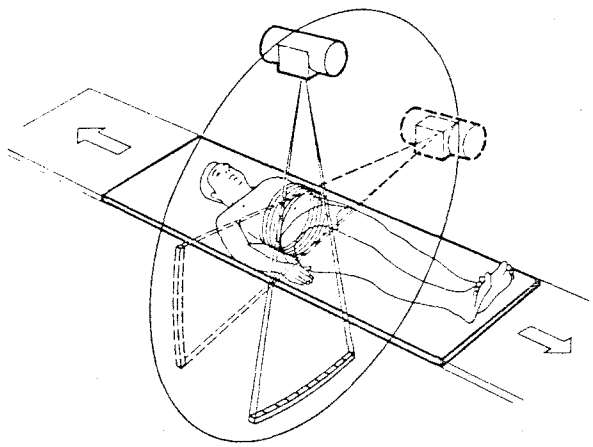
CT is a special radiography technique, which is based on the conventional X-ray diagnostic techniques and computer methods. It uses the density differences of various tissues in the human body, that means it depends on a single parameter.

In CT a great number of images obtained from different directions are processed together by a computer and a three dimensional image is calculated. So there is not any loss of information in means of dimensions.

CT examinations usually take a longer time wrt conventional X-ray. In a CT unit the tube rotates during the examination around the lying patient and the X-ray signal which penetrates and then leaves the

patient is received by a detector system, which is either stationary or rotates synchronously with the X-ray tube. The information coming from the detectors are evaluated in a computer. Using reconstruction techniques which are similar to the MR imaging methods C T images are obtained. The basic construction of a CT system is explained in the following figure.

Fig.6.4. Basic structure of a CT system



We can say that CT is a diagnostic imaging technique which can be compared with MR since both of these modalities provide computer reconstructed three dimensional images and can give similar diagnostic information. Therefore, in the following sections comparisons from the medical point of view will be made between MR and CT only.

6.2. MR versus CT

6.2.1. Head

Neuroradiological applications are of great importance with MR, and imaging the head alone could keep any magnetic resonance scanner busy and perhaps pay its way, because MR in this organ reveals what other exams may not.

Head images are also among the most visually striking. In fact, they have become emblematic of MR's superior contrast resolution. Most MR examinations of the head so far have focused on the brain,because MRI's sensitivity to brain tissue is much better than CT's. There are no bone artifacts. Respiratory motion is not a problem as it is not in CT.(12)

MR's better contrast resolution is evidenced by better differentiation of white and gray matter, more frequent detection of metastases, multiple sclerosis plaques and other demyelinating diseases, and earlier detection of infarcts and edema.

Wilson's disease affects the basal ganglia. Occasionally visualized with CT as areas of decreased attenuation coefficient in the basal ganglia, it is more frequently and clearly visualized with MR as areas of increased T1 and T2 relaxation rates.

With CT acute hemorrhages usually have increased attenuation coefficients. However, the blood sometimes

becomes isodense within the brain, resulting in decreased attenuation coefficients. On MR, with different pulse sequences, blood can be seen in all of its phases, even when isodense with the brain on the CT scan.

With CT, it is more difficult, in some cases, to differentiate calcification and hemorrhage. With MR, hemorrhage is visualized as a high intensity on T2-weighted images and calcification as a low intensity on both T1 and T2-weighted images.

Although less sensitive than CT, MR is more specific for differentiating lesions. Because of increased water, almost all pathological conditions have a prolonged T1 relaxation rate and are darker on T1-weighted images. The increased water may be secondary to the breakdown of the blood-brain barrier in some acute conditions, such as infarction, and in such infarctions as cerebritis, meningitis, abscess, and primary and metastatic malignancy. The same conditions can also be seen in the post-surgical state.

With MR, rapidly flowing blood, aneuryms and arteriovenous malformations give a decreased signal on both T1 and T2-weighted images.

The only lesion that have been shown to cause shortened T1 relaxation rates are benign tumors and some lesions containing fat. If the identification of pathologic lesions using calculated T1 and T2 relaxation times can be realized, this will be a great step in diagnostic medicine.(103)

Theoretically, lesions of the brain stem and foramen magnum can be better visualized with MR than with CT. In practice, however, some patients with demyelinating diseases, infarcts and tumors may not be able to remain motionless enough to obtain an adequate examination with MR.

For eye studies, the resolution of MR with standard receiver coils is much poorer than CT. But when special surface coils are used with a high magnetic field strength system, MR's contrast and resolution are superior. MR can image patients with sella turcica in the axial, coronal and sagittal planes without reformatting the images. With CT, direct scanning in the coronal plane can only be obtained in patients with supple necks. Frequently, in the coronal plane, CT images are degraded by artifacts from dental fillings, a problem that MRI does not have. It is also much easier and more useful to image in the sagittal plane with MR than with CT. The cella turcica lies deep within the skull to be examined with MR using surface coils. It is also not clear yet if the superior contrast resolution of MR will compensate for its decreased spatial resolution compared to CT for examination of the pituitary gland.

For ear examinations, MR should become the method of choice for acoustic neuromas, because normal cortical bone does not give a signal, and these tumors can be seen not only in the cerebellopontine angle cisterns, but also when they extend into the internal auditory canals. To obtain comparable data from CT, air must be introduced into the subarachnoid space.

There are some studies which are carried out in various clinics in accordance with examinations of the head using MR. J.Pennock, M.Bydder, and R.Steiner from Hammersmith Hospital,London studied the myelinisation in the brain of 76 babys. The formation of the gray matter and the appearance of anomalies could be well observed with MRI. The gray matter could be differentiated from the white matter after the 9. month of the baby's life. The same team was able to recognize 22 malignant tumors inside of the posterior fossa with MR.(36)

C.Mills et al. from UCSF reported about different measurement modalities of MR. According to their reports, the examination can be done with spin echo mode on five different time parameter settings, and with inversion recovery mode on two. Together with the various possibilities of scan direction settings, 42 different scan methods can be applied on each patient. The best contrasts between white and gray matter could be achieved with a pulse repetition time of 2 seconds.

Examples of images taken from the head with MR and CT are shown in the figure below.
Fig.6.5a. MR image of the head of an 16-year old physically retarded boy with a several years history of insufficient pituitary(Prof.Dr.Huk,Univ.of Erlangen)(64)

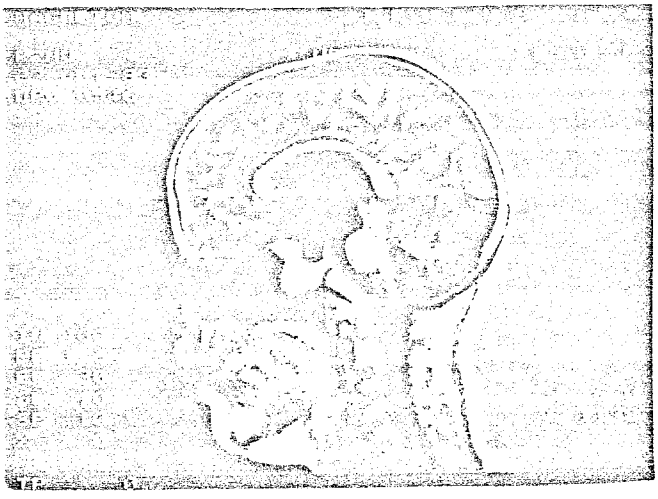
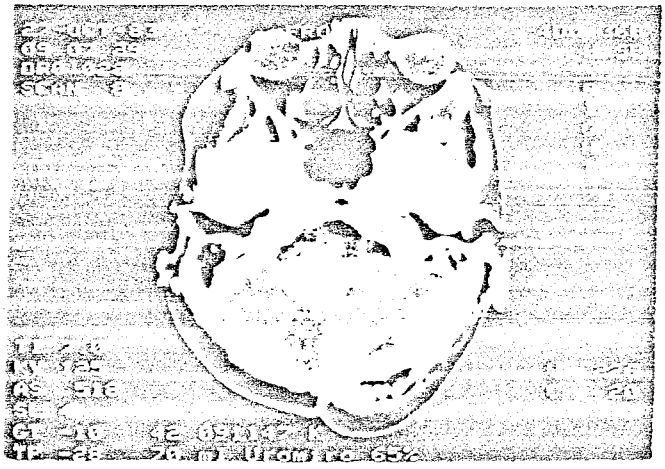


Fig.6.5b. CT image of the same patient



6.2.2. Spine

Magnetic Resonance is more accurate in detecting abnormalities of the spine than conventional X-ray films, CT with or without contrast material or myelography. MR is particularly useful in examining patients with back pain and sciatica, in part because it can differentiate normal and degenerated disks non-invasively.

With a T2-weighted image,cerebrospinal fluid(DSF) appears black (low intensity) and the spinal cord appears white (high intensity). Because cerebrospinal fluids T2 image intensity is higher than that of surrounding structures, any impingement upon the subarachnoid space is revealed. In most cases, with CT it is not possible in most cases to differentiate the

contents of the dural sac without the use of contrast material.

The normal nucleus pulpolus contains approximately 85% water at birth, the normal annulus fibrosis 80%. The water content of both decreases to 70% as the nucleus pulposus degenerates. A T2 weighted MR image can clearly differentiate the normal nucleus pulposus from the surrounding annulus fibrosis. With increased age and disk degeneration, the nucleus pulposus changes from a viscous structure to a dessicated fibrous mass. This is accompanied by loss of water and alteration in the collagenous and noncollagenous protein which can be seen in the T2 weighted MR image.(103)

In the normal spine on transverse and parasasgittal MR images, the epidural fat surrounding the nerve roots and neural foramina is well visualized. With a T2 weighted image, the individual lumbar nerve roots sometimes can be identified as filling defects within the high intensity cerebrospinal fluid.

In the evaluation of degenerative disease of the spine, magnetic resonance supplies useful information and can replace myelography in some cases. As the quality of the MR images improves, and when thinner sections can be obtained by using magnets of higher field strengths, correlative studies should be done to determine how far MR can be used in place of CT and myelography in studies of degenerative spine diseases.

Metastatic or primary neoplasms involving the spinal column can be appreciated with MR because of

changes identified in the T1 and T2 relaxation rates. In most cases neoplasis shows a decreased signal intensity on the T1 weighted image, indicating a long T1 and an increased signal intensity on the T2 weighted image, indicating a long T2.

Magnetic Resonance is a very sensitive modality for the evaluation of disk space infections. The infected disk space demonstrates a homogeneous, decreased signal intensity of the contiguous end plates. Affected intervertebral disks on a T1 weighted and contiguous plates on a T2 weighted image have an increased signal intensity.

Tumors involving the vertebral bodies have long T1 and T2 times. These tumors are not as symmetrically centered on the interspace with involvement of the disk as it is at disk space infection. MR is positive at an earlier stage than CT or than plain films and appears to be more specific than radionuclide studies for disk space infection. Some examples about the MR examinations of the spine is shown in Figure 6.6.

Fig.6.6. The schwannoma of the Cervical Cord is demonstrated in sagittal(a) and Coronal orientations(Prof.Dr.Gerstenbrand,Univ.of Innsbruck) (64)

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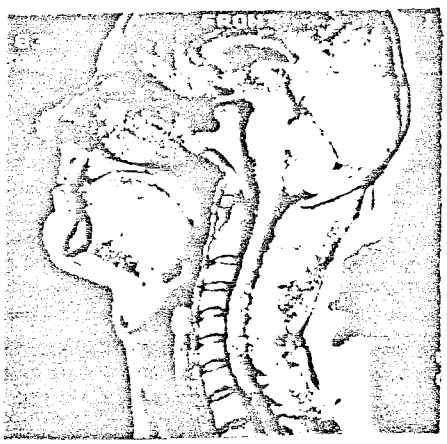
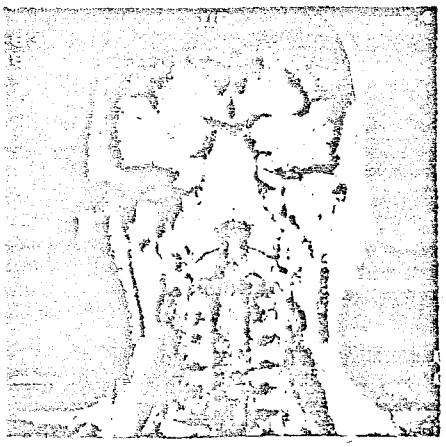
a.



b.



6.2.3. Heart and Lungs

High quality cardiac images showing anotomic detail are produced with cardiac gating. ECG triggered images are taken in routine examinations, the breath controlled imaging technique, however, is not yet well developed.

Applications of the cardiac MRI include measurements of blood flow, tissue characterization by relaxation times and metabolic tissue characterization by chemical shift.

Differences in blood flow velocity are indicated by varying intensities of the MR signal. About the applications of MR imaging, there is a study of J.Singer, from University of California, Berkeley. In his work he showed cross-sectional images of the veins of the neck, in which the proton resonance signal was proportional to the flow velocity. To achieve this, the protor magnetization was first depolarized by an RF pulse in a selected section. When after a given time, this layer is represented by an imaging spin echo series, only that portion of the blood can provide an MR signal, of which the protons hadn't been depolarized by the prior pulse. Thus the signal strength will be proportional to the quantity of protons that have flowed in between the RF pulses.

D.A.Feinberg from San Fransisco described the pulse triggering in the blood flow measurement in the

arteries. He suggested following the flow perpendicularly to the represented plane with the aid of a special 90° -180°-180° pulse scheme in three successive planes. Through the use of the field gradient in the direction of flow, a phase coding of the moving spin is effected, from which the flow velocity reciprocal to the coding duration can be read out. (36)

Until the introduction of cardiac gating, it was possible to obtain images providing good visualization of large and medium size blood vessels; the imaging of the heart itself was suboptimal, however, because of the blurring effect of cardiac and respiratory motion.

There are a variety of methods with which MRI of the heart can be gated. Most commonly, cardiac gating is synchronized to the R-waves by ECG gating. An ECG amplifier detects the ECG signal within the R-R interval, a telemetry unit transmitter sends the signal to a receiver. Computer software enables the operator to vary the delay after the R-wave in order to select a specific time within the R-R interval to trigger the RF pulse sequence and begin data acquisition.

The effects of respiratory motion has been solved by simultaneous cardiac and respiratory gating, where a variable resistive harness around the chest detects pressure changes during the respiratory cycle. The slowly changing DC signal triggers a gate so that data are recorded either when the voltage rises above or falls below a preset level. The expiratory phase which is longer, is usually chosen and the ECG signals that

fall within this phase are used during data acquisition. The marked contrast between the low intensity signal of flowing blood and the higher one of the myocardium and blood vessel walls is an important advantage of MR over CT.

In a gated cardiac MR image, flowing blood appears black, because excited nuclei disappear as the blood flows out of the gradient field during the RF pulse sequence. The blood vessel wall and myocardium appear grey to white depending on the amount of signal they emit.

As in other organs, MR's ability to obtain data directly in the axial, coronal and sagittal planes without the image degradiation of reconstruction technique of CT is a great advantage of MR.

As reported by P.Lanzer,UCSF, on a study concerning 35 patients with cardiac disease, myocardial infarcts were clearly detectable with ECG triggered spin-echo images. For the measurement, the spin-echo read out time was prolonged to 56 msec.(37)

N.Wojtowycz and E.Zeitler from Erlangen,Germany showed in the Meeting of the Society of MR, Chicago ECG-triggered images of the heart and also representations of the mediastinum and lungs with and without pathological lesions. The MR scans belonging to this study were obtained at a 0.38T device. They revealed a wealth of detail, among other things hilar lymph nodes were distinguishable from the hilar blood vessels and the mediastinum.(36)

According to F.Steiner, Hammersmith Hospital, London the ability of MR techniques to detect tumors in the hilus of the lungs is especially important. In his studies he measured lengthened T1 and T2 relaxation times in all mediastinal tumors. On the basis of their greater proton density, tumors of the lungs were easy to localize by magnetic resonance images.

6.2.4. Abdominal Organs

Extensively parenchymatous organs, such as the liver, spleen and kidneys ought to be particularly susceptible to representation using proton spin tomography, since with its aid, small changes in water and fat distribution and in the bindings state of the protons can be detected. The optimization of the scanning parameters is, however, still a problem.

N.Rupp, M.Reiser, and E.Stetter from Institute of Radiology, Munich made a study on 44 patients in the abdominal region. They prepared two tables as a result of their study which can help obtaining high quality MR images and interpreting them in a correct way, as shown in Tables 6.1 and 6.2. (82)

Table 6.1 Relaxation times in normal and pathological tissues

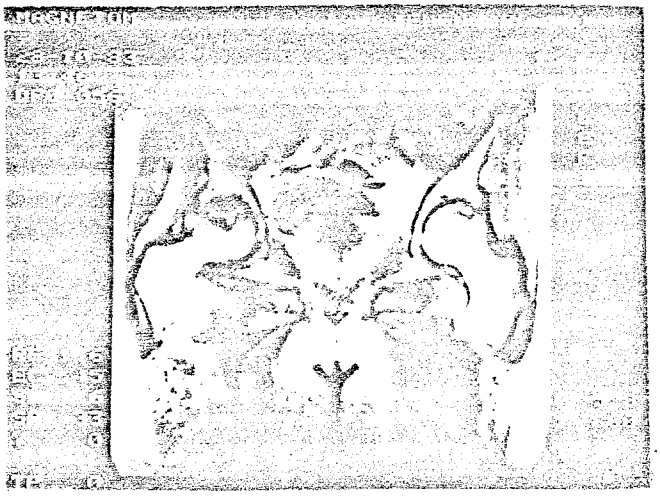
	Normal		Pathological	
	T1 (msec)	T2	T1 (msec)	T2
Pancreas	290	60	240	40
Liver	380	40	570	40
Kidney	670	50	570	110
Prostata	2200	300	610	140

Table 6.2 Signal intensities of various tissues

	T1 weighted pictr.	T2 weighted pictr.
High intensity (light)	Fat	Fluids
	Bones	
(grey)	Pancreas	Fat
	Liver	Bones
	Muscle	Kidney
	Cortex	Liver
Low intensity (dark)	Vessels	Vessels
	Fluids	Air
	Air	

Also in the pelvic area the absence of any radiation exposure, in combination with the good representation of soft parts makes MRI highly suitable. Unlike studies in the upper abdomen, there is little image degradiation due to respiratory motion. Pelvic anatomy is less complex than abdominal anatomy. The fat planes are more abundant, symmetrical and predictable. Bladder urine and rectal air are natural MR contrast agents. These, combined with wellvisualized fixed muscle planes provide high contrast definition of the pelvic organs. Figures 6.7 give an example for MRI of abdominal and pelvic regions.

Fig.6.7 Femoral Head Ischemic Necrosis of a 60 year old man with one year history of hip pain after a dashboerd injury (Prof.Dr.Lissner,Radiology Clinic-University of Munich) (64)

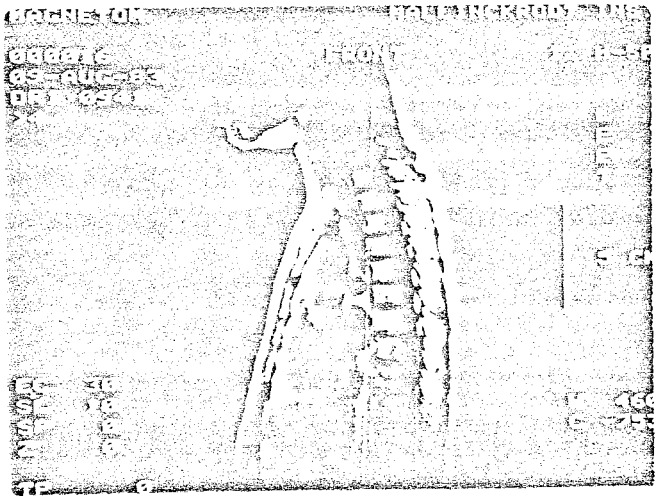


6.2.5. Skeleton

The structures of the skeleton and the surrounding soft tissues can be easily differentiated by means of an MR image. The compact bone appears dark due to the absence of mobile protons. The fat tissue is at the other end of the gray-value scale and gives very little light images. Muscles, bonds and cartilages are acccrding to their signal intensity between these differentiable limits.

A specially important parameter for pathological changes in the skeleton is the change in the signal intensity in the marrow. In the case of a hip-bone necrose, for example, MRI was the only imaging method, that lead to the diagnosis, where CT also couldn't give a detail rich image. In Figure 6.8 an MR image of skeleton is given.

Fig.6.8 A slightly atrophic tethered spinal cord is demostrated (Mallinckrodt Institute of Radiology, Washington)



6.3. Summary

The gray scale of MRI which represents the intensity of the resonance signal depends on the number as well as the chemical combination of protons, of which the latter is responsible for the relaxation times T1 and T2. Together with motion of protons they are the main factors of tissue contrast. With these parameters MRI has proved to be a very sensitive method for lesion detection, superior to the CT in many cases, such as certain brain tumors or demyelinating disease (multiple sclerosis, leucodystrophy, brain maturation etc.), without the hazards of ionizing radiation. However, small calcifications which are important for the preoperative diagnosis of brain tumors can be missed.

In fact, MR is preferable to the other diagnosis methods, because it can distinguish malignant and benign tumors, thereby revealing their nature. Calculated T1 and T2 relaxation times are more indicative of whether a tumor is benign or malignant than the attenuation coefficients of cranial CT.

The rate of blood flow in vessels can be determined by altering MR pulse sequences. By varying these sequences, MR can be used to differentiate tumors, arteriovenous malformations and aneurysms.

MR may be useful in identifying and characterizing types of fat to determine if they are saturated,

unsaturated, long chain, or short chain. In CT all fats have similar attenuation coefficients.

For topographic information, which is essential for the surgeon the direct imaging of frontal and sagittal sections is of great advantage. This is true for lesions in the head but even more so for the spine where myelography without contrast medium becommes possible. With better spatial resolution and contiguous sectional imaging these possibilities will make MRI the prccedure of choice in the examination of the spinal cord.

One of the most important advantages of MR is the deperdance of this modality on more than one parameter contrary to the other imaging methods. For instance, CT and radiography depend only on the tissue attenuation of radiation. Ultrasonography depends only on the reflection of sound waves. MRI is more complex and provides more information because it depends on multiple parameters. Additionally, multiple techniques are available for data acquisation, including spin echo, inversion recovery and free induction decay. Beside this there are variations in pulse interval, time of spin echo, 2- or 3-dimensional reconstruction of the data, type of matrix, number of slices per scan, thickness of slice etc.

All these combinations and permutations of techniques make it possible to modify the method of data acquisition until the optimal image is obtained.

MR also have some disadvantages in clinical applications, such as the slowness of scanning time, the expense and the fact that approximately 20% of patients in an acute hospital setting are not suitable to be examined by this method. For instance, MRI cannot be used in patients who have pacemakers, arrythmias, or have recently introduced metallic clips which occlude vessels. Patients that require continuous nursing care during the examination period are not suitable subjects for MRI. Some patients can be bothered by clasutrapho-bia while they are in the scanner due to its design.

The long scan time was formerly an additional bothering effect for the patients. But with developed softwares the scan time decreased to an average value of 20 minutes for an MR examination. For CT this time is not less than 15 minutes.

6.4. Contrast Agents

Paramagnetic agents may extend the diagnostic potential of MRI. Initial investigations have focused on paramagnetic ions(manganese Mn ,gadolinium Gd ,etc.), stable free radicals(pyrolidine and piperidine derivate), and molecular oxygen O_2 . These substances decrease T1 and T2 with a subsequent increase in signal intensity on MR imaging. Promising studies and initial results were reported on gadolinium, a rare earth element with the highest paramagnetic moment.

At the present time, however, contrast media for clinical use in the MRI routine diagnosis are not yet

available.

6.5. Possible Side Effects of MR

As many biological functions are based on electromagnetic interactions, three sources of possible harmful effects on health have to be considered: static magnetic fields, changing magnetic fields, and radio frequency heating. Harmful effects on humans and reproducible cellular,biochemical,or genetic effects have not been observed at static magnetic fields less than 2T.

6.5.1. RF Heating

The major effect of RF fields on the human body is heating. The safety level established by the United States Occupational Safety and Health Administration is 10 mW/cm^2 incident electromagnetic power density for 10 MHz to 100 GHz averaged over a 6 minutes period, which is based not on what man can tolerate but on the concept that the electromagnetically induced heat load should not exceed the basal metabolic rate of man.(13)

MR RF power systems deliver incident power densities below the range of this standard. However, RF frequencies near 70 MHz will produce significant heating problems for power densities near 10 mW/cm^2. The energy absorbed by man varies with frequency, orientation and organ. Resonance for man is about 70MHz and at frequencies near resonance, the maximum power deposition effect occurs at the point where the wave

first strikes the body.

Thus, in designing MR instrumentation, it is prudent not to exceed 10 mw/cm^2 average power density, although twice this power density should be acceptable for exposure periods of less than 30 minutes.

6.5.2. Static Magnetic Field Effects

There is no substantial biological evidence reported for static magnetic field effects for fields below 2.0 T. Fields of higher strength show influences on cell morphology and on enzym reactions in test tube studies, but no such effects have been reported for subjects who have spent considerable time in magnetic fields up to 20kG in nuclear physics laboratories.

6.5.3. Time Varying Fields

Time varying magnetic field gradients which are applied during the measuring sequence induce weak electrical currents in the human body. At frequencies in the range of 10 to 100 Hz there have been reports of a sensation of light, so called phosphenes. Harmfull effects to the central nervous system, however, have not been observed.

In summary, no permanent side effects of MRI on organisms have been described at the present time. Further epidemiological studies are necessary to reveal possible hazards yet unknown.

7.INSTALLING AN MR-SYSTEM

The placement of a Magnetic Resonance System withinthe hospital environment causes some difficulties and problems,which do not appear to that amount for other types of medical equipments.Architectural requirements are among the first issues that must be taken into consideration by a facility planning an MR installation.Specifically, the system's structural, mechanical,electrical,and spatial requirements must be evaluated.

Certain fundamental characteristics of an MR system, most notably the strong magnetic fields and radio frequency energy utilized, impose definite constraints on an MR suite's architecture.

7.1. Basic Considerations

7.1.1. Magnetic Field

Like all magnetic fields,the static field of the MR system is three dimensional and extends to the space above and below the system,as well as to surrounding space on the same level.Magnetic field strength decreases as one moves away from the magnet's center.The field extends out further in the direction of the bore axis.

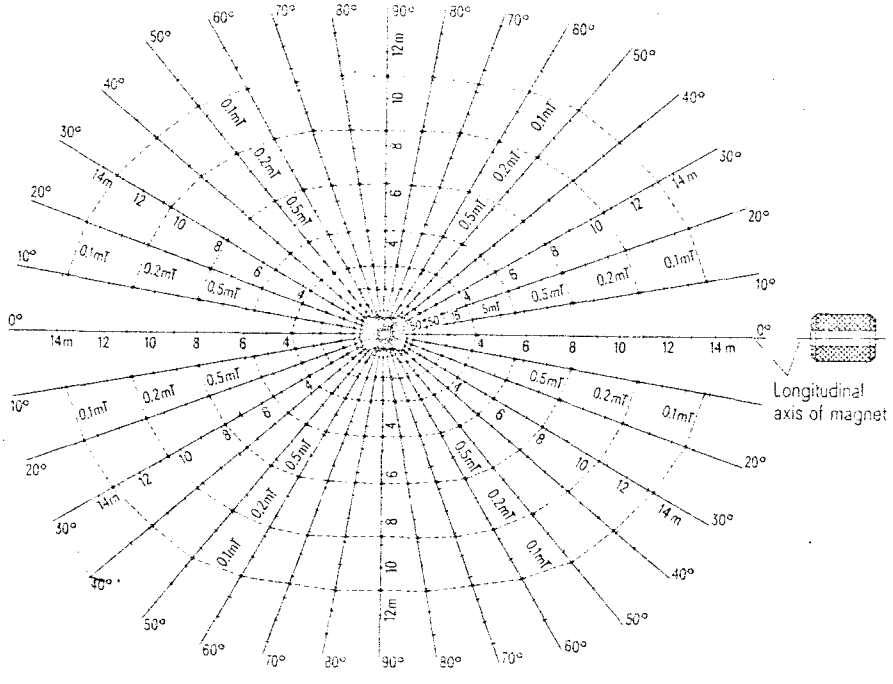
This large three-dimensional field is the major limiting factor in system location and architectural

design.The basic problems posed by the magnetic field are two fold.First,ferromagnetic materials in the environment can distort the magnetic field of the system,degrading image quality.Second,the magnetic field can interfere with the function of certain mechanical or electrical devices such as nuclear cameras and pacemakers.Site selection and architectural plans must minimize or eliminate these problems.

7.1.1.1. Influence of the Magnetic Field on the Environment

For room planning,it is important to know that the so called stray magnetic field spreads out three dimensionally around the magnet.Therefore,the storeys above and below must also be taken into consideration in the planning.To illustrate how the stray field spreads out, Fig.7.1 shows the measured field distribution of a 0.5T superconducting magnet in one plane.(68)

Fig.7.1 Distribution of the magnetic field of a 0.5T magnet in air



It can be seen from the illustration that the distance required for the field strength to drop to the level of the natural magnetic field of the earth is 17m in the direction of the magnet axis and 14m perpendicular to it. The field distribution in Fig.7.1 was measured in the air. However, if the magnet is in a building with heavy steel reinforcing, e.g.concrete costruction, then according to theoretical calculation,the true relationships can deviate by the factor 2 from the ideal curves and interference can occur in units at an even greater distance than those listed in Table 7.1.

Table 7.1 Examples for the influence of the magnetic field on units in the vicinity of the magnet

Magnetic Flux Density	Minimum distances which must be maintained from the center of the magnet(x/z axis) 0.5T	1.5T	Units affected
3mT	3.5/4.5m	5.0/6.2m	Small motors, clocks,credit cards, magnetic disks,photo equipments
1mT	5.0/6.5m	7.0/9.0m	TV installations, image display screen, disk drives
0.5mT	6.5/8.0m	9.0/11.5m	Cardiac pacemakers
0.1mT	10.5/13.5m	15.5/19.5m	Image intensifiers,CT, Gamma cameras

The most important unit which must be taken into consideration in clinical operation is the cardiac pacemaker.The influence is due to a magnetic switch used in pacemakers. Persons with cardiac pacemakers are not allowed to be exposed to magnetic flux densities greater than 0.5mT,since this gives rise to abnormal functions.

By taking a certain safety factor into consideration for the limit values,the types of units which can be influenced by magnetic fields can be allocated into four groups of maximum permissible field strengths. Broadly considered,the groups are:
units with mechanically sensitive parts,such as clocks and photographic apparatus;a further group includes magnetic storage devices such as magnetic disks and magnetic tapes,as well as credit cards,even computers to a certain extent react sensitively to magnetic fields. A large group of interference-sensitive units are those in which electrons move in a vacuum.This includes TV monitors, oscillograph tubes,x-ray image intensifiers and gamma cameras. Even CT units can suffer interference from magnetic fields.

7.1.1.2. Influence of the Environment on the Magnetic Field

MR systems anticipated applications,which affect the type and size of magnet used and the attendant environmental requirements is the key determinant of basic site selection. If it is decided to confine MR applications to hydrogen imaging,the field strength

and homogeneity requirements of the magnet will not be as stringent as those necessary for chemical shift spectroscopy.

In general,the greater the requirement for maintaining magnetic field homogeneity,the more strictly the MR site must conform to the ideal non-ferrous environment. For hydrogen imaging,magnetic field inhomogeneities will distort the position information in the scan volume,reducing the image quality. To prevent this problem,it is anticipated that a magnetic field homogeneity of between a maximum of 100 ppm and a minimum of 10 ppm will be required.Generally,higher magnetic fields and more stringent resolution requirements will dictate higher magnetic field homogeneity.

For obtaining a high homogeneity,following sources of interference should be examined:

a-) Distribution of iron-steel in the building,
b-) Moving iron objects and fields,
c-) Slow AC fields.

7.1.1.2.1. Distribution of Iron-Steel in the Building

The existence of steel in the building construction distorts the stray field of the magnet and thus reduces the homogeneity of its field within the volume to be measured.

This static influence may be corrected up to a certain steel concentration by additional coils in the magnet,so-called "shim coils".Understandably,the distance of any particular steel reinforcement from the magnet is important.A check whether the homogeneity correction is possible by shims,can be made by a calculation of the iron-steel distribution in the floor walls,ceiling,support beams and columns. As an example, the floor beneath the magnet should not contain more than 15kg/m2 structural steel,although it must carry a weight of more than 5 t.This is best achieved, if a concrete pedestal with minimal steel reinforcement is cast onto solid ground.

7.1.1.2.2. Moving Iron Objects and Fields due to switched on DC currents

If objects,magnetized in the magnetic field of the earth or in the stray field of the MR unit,move in the neighbourhood of such a Magnetic Resonance System,then the magnetic field in the volume to be measured is also influenced.The degree of this influence depends principally on the shape and size of these objects and the distance form the magnet.Furthermore,it is not possible to exactly determine this influence before the installation of the MR unit.

In order to check the suitability of a desired site,it is therefore important to know,whether,for example,the influences of a lift,the movement of a transport vehicle in the passage way or perhaps the

vehicles on a neighbouring street can be tolerated.An investigation in which small field changes were simulated by a computer gave the result that in the case of proton tomography,flux density changes of 0.1 mT were just permisseble. With greater field fluctuations,artefacts must be reckoned with in the image reconstruction.As an example,investigations with an automobile gave the result that with a 0.5 T MR system at a distance of 12m the magnitude of the magnetic interference field changes are permissible.

Attention should also be drawn to magnetic fields in the vicinity due to switched on DC currents which also must be regarded as interference fields.DC-driven trams and underground trains could,for example,generate such fields that interference effects occur frequently at distances of 30m to 50m. Cases can also be imagined, in which it is not the vehicle itself,but the earthing currents in the steel reinforcing of the building which causes interference to the magnetic field.

7.1.1.2.3. Slow AC fields

Slowly changing magnetic AC fields(e.g.50Hz power line frequency)can influence the magnetic field. Since the screening of such fields is not easy to carry out, there remains only the possibility of maintaining distances to such sources of interference as large as possible,so that the interference fields to be expected are still tolerable for the imaging.Transformers and high-power cables in the vicinity of an MR device must be regarded as potential sources of interference,which

can be permitted up to 0.1 mT or slightly higher than this value. Examples of influences from interference generating units or structural elements in the vicinity of the magnetic field are shown in Table 7.2.

Table 7.2 Examples of influences from environment on the magnetic field

Interference generating units or structural elements	Minimum distances to the mid-point of the magnet
Reinforcing steel meshes in the floor	1 m
Steel beams,supports, reinforced columns	5 m
Wheel chairs	8 m
High power cables,transformers	10 m
Automobiles	12 m
Lifts,trucks	15 m
Railways,trams	30 m

7.1.2. Radiofrequncy Fields

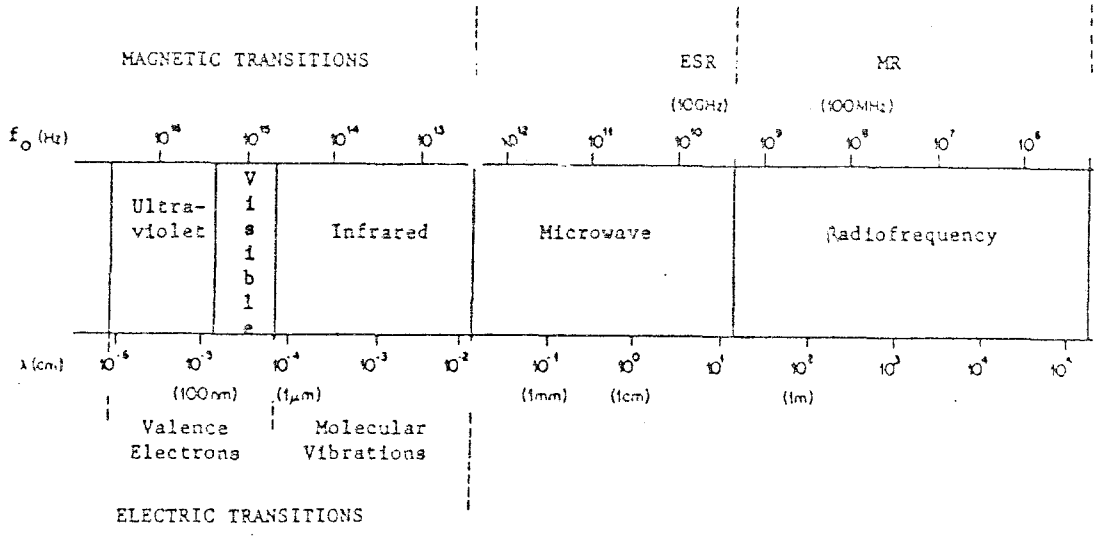
MR examinations require the reception of extremely weak radiofrequency(RF) signals from the patient's body. It is therefore obvious that external sources of interference,above all those of the same frequency as the MR signal coming from the body,falsify the measured values received,reduce the image quality and can generate artefacts in the image.It is true that a measurement of the RF field strength and its spectrum

can determine the momentary situation,but this offers no security against future interference,since the reception field strength is in the short wave range, and it depends on propagation conditions. Even transmitter can come into operation.

The operational frequency for all MR units lies between the AM and FM portions of the electromagnetic spectrum. This is shown in Fig 7.2. In the bottom of that figure are the corresponding magnetic field values for specific frequencies.In this frequency range numerous broadcast stations and radio operations are found.In addition white RF noise is emitted from fluorescent lights and discharging capacitors of passing motor vehicles can interfere with MR signal. Since the user can exercise no influence on such sources of interference, RF shielding is necessary.

A further aspect which in part forces the provision of RF screening is that an MR installation functions as an RF transmitter as long as the RF coil is switched on and this can cause interference in sensitive units in its vicinity.

Fig.7.2 Frequency Allocations(Log Scale)



7.2. Site Planning for an MR System

For eliminating or at least reducing the problems mentioned in the previous sections,a very carefully site planning has to be done before the installation, considering following points:

a. Magnetic Field Shielding,
b. RF Shielding,
c. New Building or Renovation.

7.2.1. Magnetic Field Shielding

Magnetic field shielding presents one of the most difficult site planning problems,and this is a strong function of the size of the MR imaging system selected and its location.There is a distinction between the shieldings according to the type and strength of the magnet.(75)

7.2.1.1. Permanent Magnets

Below 0.3T,the advantages of a permanent magnet are that less space is required so the magnet can be readily installed in relatively small sites;and no magnetic shielding is required.A major disadvantage of the permanent magnet is its weight of about 100 tons.

7.2.1.2. Resistive Magnets

Resistive magnets may reach 0.25 T through special windings and integral shielding,and they have the advantage that they can be turned on and off daily.On the other hand they have large power and cooling requirements,so that installation brings some problems.

7.2.1.3. Superconducting Magnets

For this type of magnets,and especially for high fields magnetic field shielding is required unless the site is unusually isolated.A method of shielding, which is called as ' mirror shielding ' is to use external carbon steel plates with a high permeability.

Depending upon the site,the magnet may need to have the shielding on all six sides.The enormous weight of such screening and the tremendous forces arising between the magnetic coil and the iron screening is, however,a disadvantage.

If for example,a single x-ray installation with image intensifier is in an area where the field strength is only slightly higher than that of the earth field,then a room screening of the image intensifier system seems to be a better proposition than screening the complete MR system.In this case,screening material of a few millimeters thickness is adequate. Which screening material is the best must be determined for each individual project.

As already mentioned, wearers of cardiac pacemakers must not enter an area where the magnetic flux density is greater than 0.5 mT. If, as shown in Fig.7.3a, this area overlaps partly into a passage way outside of the MR department, then the field strength must be held back by appropriate screening. Fig.7.3b shows the field curve obtained by means of screening which allows cardiac pacemaker wearers to pass by without danger.

Fig.7.3a 0.5mT limit of the magnetic field without any screening

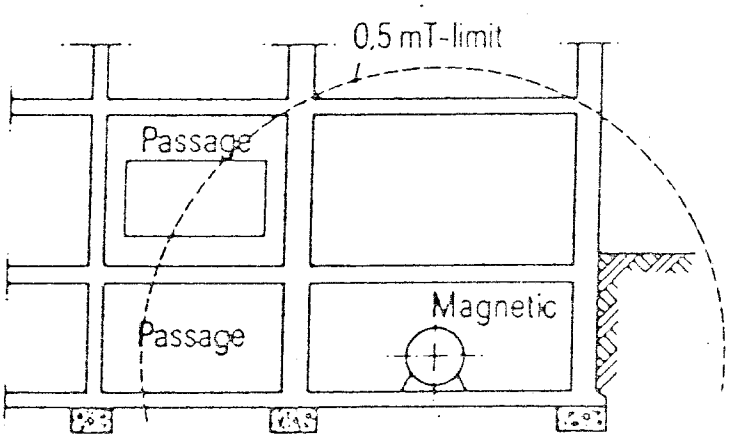
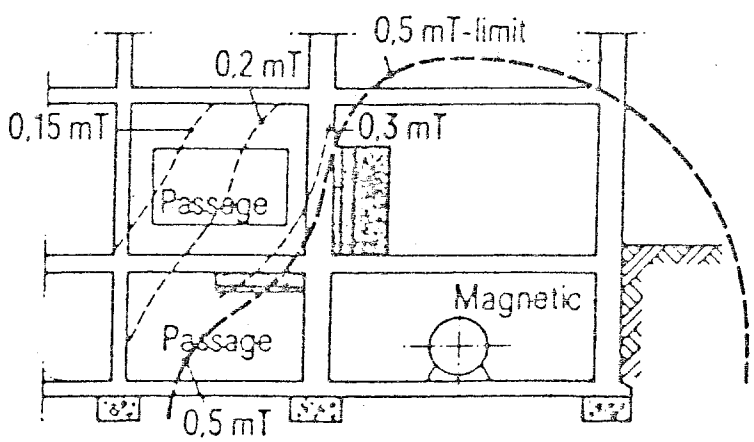


Fig.7.3b 0.5mT limit after fitting suitable iron screening



7.2.1.4. Magnetic Self Shielding

There is a fairly new method,which is called Magnetic Self-Shielding and is introduced by a few manufacturers. It offers the following advantages:

- Drastic reduction of the distances of the stray magnetic field lines in all three dimensions,

- Installed in the magnet housing,

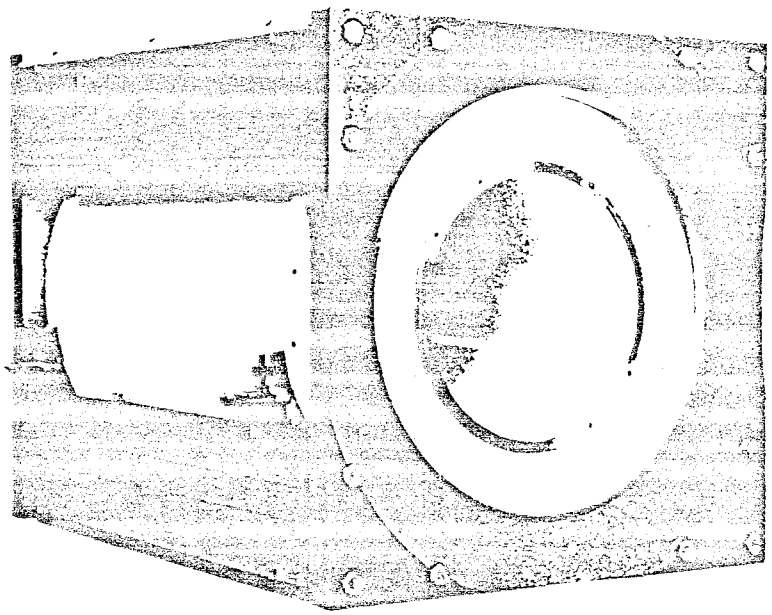
- Reduction of the required floor space within the MR department,

- Eliminates the need for extensive magnet room shielding at the walls,ceiling or floor,

- Offers more flexibility in selecting an installation site.

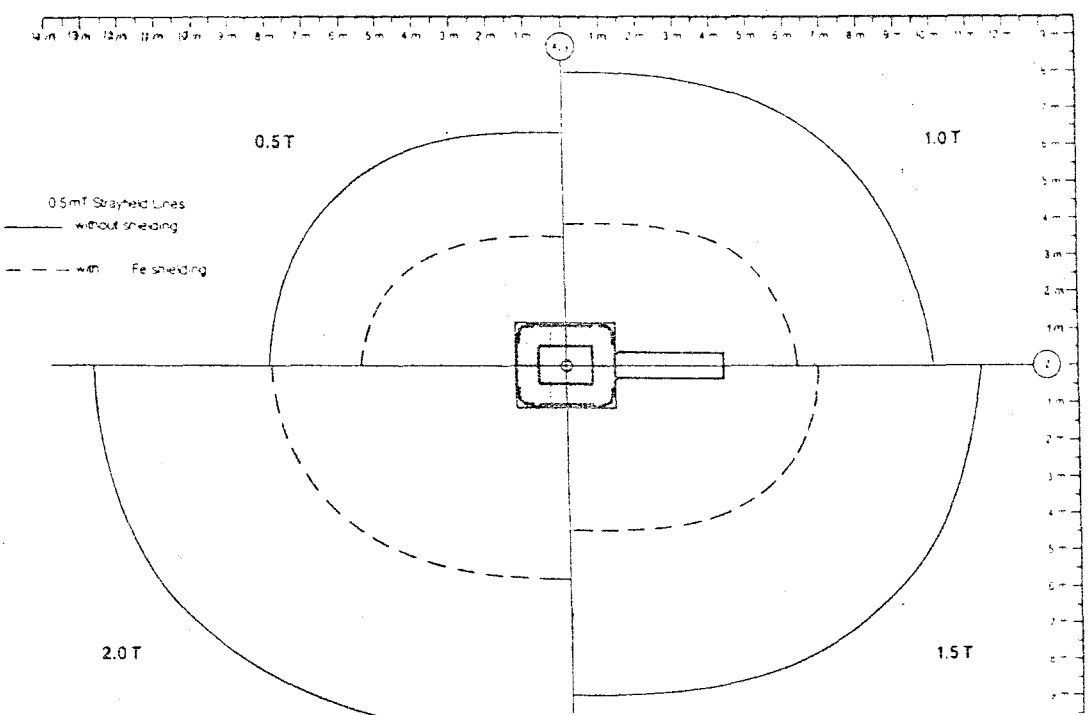
Magnetic Self-Shielding consists of two face plates and four sets of beams parallel to the magnetic axis as seen in Fig.7.4. Optimal shielding is achieved by selection of proper material,shape and special treatment of the surfaces.The magnet can be adjusted mechanically with respect to the cryostat.The high homogeneity of the magnetic field will not be changed.

Fig. 7.4. MR device with self-shielding



In Fig.7.5 a comparison between shielded and not shielded MR systems of various field strengths is given

Fig.7.5 Magnetic Field Distribution of Shielded and Non-Shielded MR Systems



7.2.2. RF Shielding

The electromagnetic shielding has to handle two tasks:

- To protect the environment from interfering effects of the exciting pulse during the transmit phase and

- To prevent high frequency interference of the environment from reaching the sensitive measuring instrument of the MR tomograph during the receive phase.

Shielding of magnetic and electromagnetic fields is necessary during excitation of magnetic resonance.RF tight shielding of the room with the aid of electricall conductive metal sheets or metal lengths joint by means of soldering,welding,screws or spring strips,is absolutely necessary.Electrically conductive wallpapers or metal powder additives in plaster and floor covering are completely inadequate shielding materials and they offer at the best a slight shielding effect against electrical fields.

There is a non-ferrous modular design for room shielding.This design laminates aluminum or copper sheet to both sides of the board. The panels are fabricated in different sizes according to manufacturer.

Producing the panels is relatively simple to do,

but the joining together is the key to a good RF tight enclosure. Almost all metals can provide much more attenuation in theory than in practice; the key to high performance is not choice of metarials but a tested shielding system of good design and workmanship in handling all joints, all openings, and all discontinuities of any type. Of all such openings, the door is the most critical. The effectiveness of the entire room-regardless of size or materials of construction- can be degraded significantly by one improper opening.

Thus, a good clamping system will certainly make an RF-tight joint, but a completely sealed room is of little or no use. The basic needs of a room are electricity, may be water or a phone, and of course a door-and the most difficult item to design and manufacture is the RF door.

The RF door must be designed for reliability, for relative ease of operation, and for smooth entry into the MR suite. The knife edge design, or recessed contact mechanism (RCM), is proven to be the most reliable, the safest, and the best attenuator.

All leads coming into or going out from the examination room must pass through an appropriate wall filter for the purpose of RF-interference suppression. Also the ventilation ducts and other incoming elements such as nitrogen pipes etc. must have interference suppression. All in all, a rather expensive construction for the RF cubicle.

7.2.3. New Building Versus Renovation

The advantages of a new building versus renovation are that more space can be obtained, and B-field shielding requirements are reduced. In addition, a new building normally provides better access for outpatients, and the total operating costs of a new building are more clearly defined. Disadvantages of a new building are that inpatients may have to be transported longer distances.

Another disadvantage of a new building is that a communications system for inpatient records, reports, and images for both the consulting radiologist and referring physician needs to be developed.

An advantage of renovating an MR site inside radiology department is that radiologists will have ready access to the MR for consultations. Obviously, any structure for renovation must provide easy access for outpatients. Another advantage of such renovation is that duplication of a large number of services can be avoided compared to the situation when the MR site is located in a new building or at a site remote from the main stream of radiology. These duplicate services include outpatient waiting areas, secreterial/receptionist facilities, film processing system as well as billing staff. Of course, renovation implies the need for some excess capacity in these services.

7.3. Some Other Architectural Considerations For The Installation

In addition to the considerations in shielding problems,there are some other points which must be examined irrespective to the fact whether the site of installation is a new building or a renovation.

7.3.1. Prerequisites for the use of liquid gases

In an MR unit with a superconductive magnet,the innermost vessel contains the superconductor which must be kept always close to the absolute zero point(0deg.K) For this purpose,the cryostat is filled with liquid helium.Around this is a vessel with liquid nitrogen,and finally,a vacuum container for better isolation.

To maintain the temperature in normal operation,a maximum of 2 litres nitrogen and 1/2 litres helium evaporate per hour, i.e. inclusive of filling losses, every week approximately 450 litres nitrogen and every month, the same quantity of helium must be available.

However, for the first filling of the cryostat, about 3000 litres nitrogen and about 1500 litres helium are necessary. For planning, this means that outside the building, a site for a nitrogen storage vessel of about 1500 litres content should be found with access for a tank vehicle.

For pipeline lengths between the storage vessel and the cryostat of up to 10 meters, simple insulation is sufficient, but if the distances are larger, a vacuum-insulated pipe must be used. In comparison, the storage of helium involves too great a loss due to evaporation. Therefore, it must be delivered every month in special vessels. This means that, as far as possible, there should be no differences of level in the access way for the helium vessel weighing 460 kg and the door openings must be 1.50 m/2 m.

7.3.2. Other Requirements

There are some other points which must be considered before the installation: (88)

- Air conditioning: An environmental temperature of 22+/- 3 ^{0}C and a relative humidity of 40-60% should be provided. Filtration in the computer area must be 90% of all particles up to 10 μ.

- Water: A water supply of approximately 60-800 l/min. is necessary.

- Site construction material: A concrete slab on grade with minimum of steel reinforcement shoud be prefered for the floors. In wall and ceilings corstruction,masonry,wood or concrete with a minimum of steel reinforcement can be used.

7.4. Standards for the Installation

Until now, there is not yet any standard worldwide, that must be followed for the installation of an MR system. All the precautions and the fullfilments of the requirements are regulated by the manufacturers.

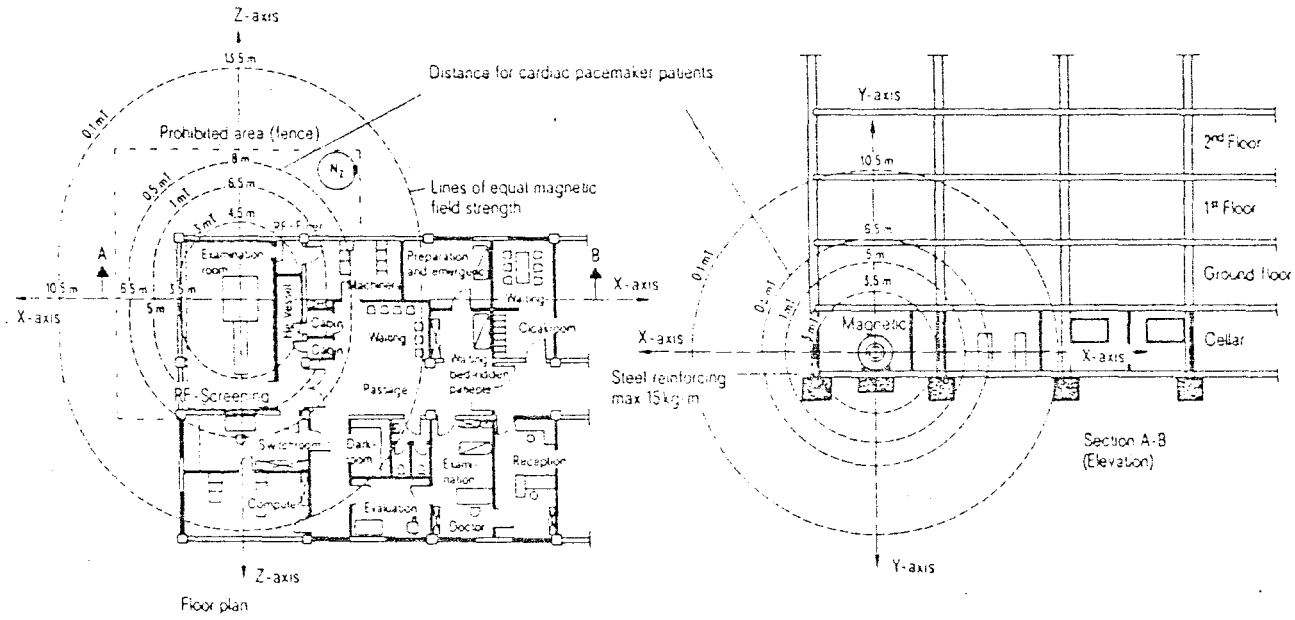
A collection of rules about the MR systems which is published by the German Ministry of Health, is one of the very few official documents about this topic. It only contains, however, recommandations to the medical staff who use MR systems, and not statutory instructions

It will surely take some more time that national or international standards about MR systems are laid down.

7.5. An Ideal Site Planning

By following all the precautions and prerequisites for the installation, a Magnetic Resonance System can be placed in a medical institution, so that it functions properly, and without causing any hazard to human beings or to other equipments in the environment. In Fig.7.6 the planning of an installation is shown, which is done according to prerequirements for a self shielded magnet in a new building.

Fig.7.6 Floor plan and sectional drawing for MR. The arrangement of the rooms is shown with their functions as well as the horizontal and vertical extent of the stray magnetic field.(68)



8. FINANCIAL ASPECTS OF A MAGNETIC RESONANCE SYSTEM

The utilization and use of any diagnostic radiologic equipment is determined by multiple factors; an important one is cost. Very often the cost can prevent health care institutions from purchasing a diagnostic system, even when the need is there.

Especially in the case of Magnetic Resonance Systems the cost plays a significant role. In determining the economic feasibility of MR for an hospital,the following breakdown of costs will be used:

- Capital costs,
- Fixed operating costs,
- Variable operating costs,
- Indirect costs.

The choice of magnet will have the greatest impact on the cost of the imaging system.In addition,it affects installation and operating costs because of space and power requirements.

8.1. Costs

8.1.1. Capital Costs

Caapital costs include the magnet,renovations of a facility,and any special heating,cooling or RF shielding.Currently available commercial MR units range from a low of $800.000 to $900.000 for a 0.15 T

resistive unit to a maximum of $2.3 million for a 2.0 T superconductive unit.At about $1.5 million are 0.3 T permanent magnet and superconductive magnets in the 0.35 to 0.5 T range.

Some facilities require expensive renovation and special ferrous shielding to protect equipment. Renovation costs have exceeded $1 million at some sites but at others can be accomplished for much less. Resistive and permanent magnet installations require less space than superconductive magnet installations,and larger superconductive magnets require more space than smaller ones.The range of space requirements is perhaps 1500 to 2500m^2 of direct floor space.Because of the fringe field,one must consider the influence on space immediately adjacent to the MR site as well as space above or below it.

Special requirements for the resistive unit include a source of chilled water-often a special cooling system that can cost more than $70.000 installed.Permanent magnets have extraordinarily weight and require a special foundation. The range of $200.000 to $450.000 for a facility renovation in most instances would be conservative. Depending on location,it is possible to spend considerably more.

To protect the room from outside RF interference, special shielding is advisable. In some instances this can be included in the purchase price of the magnet. In most instances the room must be specially built.

All of these components involve capital expenses of $1.3 million to $2.6 million. In Table 8.1,capital costs for different MR systems are shown.(4)

Table 8.1 Capital expenses

	Resistive 0.15T	Permanent 0.3T	Supercond. 0.5T	Supercond. 1.0
Magnet	900.000	1.200.000	1.000.000	1.300.000
RF shielding	50.000	included	75.000	100.000
Renovations	250.000	200.000	425.000	500.000
Total Capital Expense($)	1.200.000	1.400.000	1.500.000	1.900.000

8.1.2. Operational Costs

Operational costs are divisible into fixed costs,variable expenses,and indirect operating expenses as is shown in Table 8.2,again for four different systems.A maintanence expense of 6% of the purchase price is assumed and may be a bit high for stationary permanent magnets and a bit low for the more complicated resistive and conductive magnets.(28)

Utilities for the magnets include electrical power and for superconductive magnets,the cost of the cryogens. Power for image processing is essentially the same for all units. With a permanent magnet, the field can be generated entirely without electrical power. Resistive magnets depend exclusively on electrical power to generate the field. The annual cost of electricity varies greatly depending on the cost of electricity and the number of hours of operation of the system.

Superconductive magnets require a constant supply of liquid helium and nitrogen but have the advantage that the field is always present. Therefore, there is no additional magnet power cost to run the unit on multiple shifts per day for seven days a week. Strictly from the point of view of power consumption, facilities where extended workdays and seven-day-a-week operation were anticipated would be more inclined to consider a permanent or supercondictive magnet, whose power cost is relatively fixed whereas resistive magnet cost is proportional to hours of operation.

Personnel requirements given in Table 8.2 assume a single shift operation and staffing by a full-time physician with a strong engineering background, two full-time technicians, and a full-time receptionist/typist. Additional support (messengers, physicists, and costs of billing, bookkeeping) are included under the indirect operating expenses.

The calculation of space rental assumes a size of 1,500m^2 for a smaller facility,and a size in excess of 2,500m^2 for a larger superconductive magnet installation.

Variable expenses include the cost of film, medical supplies, paper products, magnetic discs and tapes. It appears likely that current investigation into the use of contrast materials may result in the development of injectable agents which will enhance MR images and, in the case of labeled antibodies, be specific for certain tissue types.

The total annual operation cost as detailed in Table 8.2 ranges from $534,000 to $700,000. This is based on ten examinations per day during a single shift.Multiple shift operation would increase costs, but the increased number of patients should lower the break-even point.

Table 8.2 Annual operational costs,10 MR studies per day

	Resistive 0.15T	Permanent 0.3T	Supercond. 0.5T	Supercond. 1.0T
FIXED COSTS				
Maintanence	60.000	50.000	90.000	120.000
Magnet Power	35.000	5.000	30.000	40.000
Processing power	6.000	6.000	6.000	6.000
Personnel	225.000	225.000	225.000	225.000
Rent	38.000	38.000	50.000	62.000
Insuarance	10.000	10.000	20.000	20.000
Miscellaneous	10.000	10.000	20.000	27.000
VARIABLE COSTS	50.000	50.000	50.000	50.000
INCIRECT COSTS	100.000	100.000	120.000	150.000
TOTAL ($)	534.000	494.000	611.000	700.000

8.2. Break-Even Analysis

The point at which a facility recovers all its costs is determined by a break-even analysis. This incorporates the cost of operating a facility as a function of the number of studies done per day (or per year) and from this data determines the charge per study to recover the costs.Table 8.3 summarizes the annual costs of the systems,that are discussed before.

Table 8.3 Total annual costs for ten studies per day

	Resistive 0.15T	Permanent 0.3T	Supercond. 0.5T	Supercond. 1.0 T
Capital expenses	350.000	510.000	600.000	725.000
Operational costs	534.000	494.000	611.000	700.000
Total annual costs($)	884.000	1.004.000	1.211.000	1.425.000

For the least expensive system with an annual cost of $884.000, the charge per study is almost $450 for ten patients a day. If only five patients were seen per day, the break-even charge would be about $700,but if 15 patients were seen, it would decrease to about $300. Superconductive magnets require $500 to $600 per case,with 10 patients per day.

8.3. Costs of MR versus CT

In Mallinckrot Institute of Radiology,Washington a cost analysis for a 0.5 T superconducting MR unit and a for CT unit was done. Table 8.4 shows the results obtained in this study. According to this table, the costs of MR are higher than CT by approximately 15-20% because of equipment costs and space needs.(34)

Table 8.4 Costs of MR and CT in 1983

	CT	MR
Capital costs	145.000	240.000
Personnel	261.000	290.000
Maintenance	74.000	114.000
Cryogens	—	15.000
Direct,variable	102.000	22.000
Overhead	219.000	226.000
Total	801.000	907.000

A break-even analysis is calculated with 3000 patients/year for CT and 1500 patients/year for MR. The charge is $342 for a CT patient, and $775 for an MR patient. That means, the break even point charge for MR is more than double the charge for CT. If 3000 patients per year can be studied with MR, the charge decreases to $402, or it becomes approximately 20% higher than CT.

9. MAGNETIC RESONANCE IN TURKEY ?

In the previous chapters, Magnetic Resonance as a very new diagnostic imaging method in medicine has been studied from different points of view. In that study we saw that this new method brings many advantages with respect to other imaging modalities, but also some questions about feasibility, possible hazardous effects, service problems, etc.

In this section, the positive and negative points of a possible MR installation in Turkey will be discussed. As it is known there is not yet any MR in Turkey, although more than 200 are functioning all over the world. Many considerations will be examined in order to see the chance of MR to penetrate into a turkish medical institution.

An exact feasibility calculation has to be prepared in order to decide to purchase an equipment or not. In the case of MR the feasibility depends on many factors such as the actual hospital where the MR is intended to be installed as well as the economical parameters present at the time of installation(such as inflation rate, form of payment and interest rate).

The study given in this chapter is not a complete feasibility calculation. It is only an example which can serve as a guide in preparing a more detailed realistic study for an eventual installment of an MR system.

This study is made according to the following assumptions:

- A 1.0T system with self-shielding is intended to be purchased.
-The clinic has a suitable site where MR system can be installed.
-The clinic has the necessary amount of money in order to purchase an MR system and to recover all the costs.
-The inflation rate and the interest rate are not considered in the study.
-The calculations are done for a depreciation time of 5 years.
-The exchange rate of US $ to TL is taken as 1:500.

9.1. Various Considerations

9.1.1. Medical Necessity

In order to purchase a medical equipment, its necessity must be proved at first. Medical applications of Magnetic Resonance were discussed in Chapter 6. We can make a summary of that chapter as follows and find out whether it is possible to replace the function of MR by an other imaging method:

- No other imaging technique, not even CT can challenge MR for neurological applications. With the high contrast resolution and better differentiation of white and gray matter in brain, metastases, infarcts, edema and hemorrhages can be detected in the earlier stages

using MRI.

- Through MRI benign and malignant tumors can be diagnosed without using any invasive method. Identification of different lesions seems to be realized in a near future. These two facts are not possible with other techniques.

- Using the so-called "surface coils" with high magnetic fields, superior images are obtained from certain parts of the body, including eyes, ears, and breasts.

- With MR it is possible to image patients directly in axial, coronal and sagittal planes. With imaging methods like X-ray, ultrasonography, and nuclear diagnostic devices, images only from a single plane can be obtained. CT uses axial plane images to reconstruct the coronal ones, where resolution is strongly disturbed by this process.

- For spinal cord examinations, in most cases MR replaces myelography which is a painful examination method for the patient. CT images are also poorer in detail than MR in this region of body.

- ECG triggered heart examinations give very good images because of a high contrast between the myocardium, blood vessel walls and flowing blood.

- Detailed pictures with high resolution are also obtained from other organs, including lungs, liver, spleen and kidneys.

- In contrast to other imaging methods, MR does not only depend on one parameter. There is also the possibility to make combinations of different measurement and reconstruction techniques until the best solution is found.

- MR has some disadvantages though including the slowness of scanning or the impossibility of using it with patients who carry pacemakers and metallic clips in the body.

As a conclusion to the medical necessity discussion, we can say that Magnetic Resonance is able to produce such images that it becomes a medical necessity. With an MR most abnormalities in the human body can be imaged and diagnosed. It must be added that MR should be better used in more difficult and critical cases, where CT is more advisable for routine diagnostic, since it is faster, cheaper, and more practicle than MR. That means, a medical institution should purchase a CT first and only then an MR system.

9.1.2. Possible Side Effects

Possible sources of hazardous effects on human body in MR imaging are the high magnetic field, the time varying fields and the RF waves. But no permanent side effect on living organisms caused by these sources has been described until now.

As explained in Chapter 4, MR makes use of the different distribution of hydrogen atoms in different tissues. Therefore MR is free of the proven dangerous effects of ionizing radiation which is used at X-ray and CT units. MR examination is also free of the radioactivity present in nuclear medicine imaging techniques. So far, the utilization of MR units seems to be quite safe.

9.1.3. Optimum field strength

Because of reasons explained in Section 5.1.4., the optimum field strength in an MR system for routine diagnostic is 1.0 T(85). Higher field strengths up to 2.0 T are used mostly for spectroscopic investigations.

The purchase price and the shielding requirement of a magnet with greater field strength than 1.0 T increases rapidly. Therefore, the price for the benefits brought by a stronger magnet seems very high. A magnet of 1.0 T seems to be a good choice for a medical institution in Turkey. The calculations in our case will be done accordingly.

9.1.4. Installation

There are some difficulties in the installation of an MR system which don't appear for the other imaging methods.

The high magnetic field of the MR system can

effect certain mechanical and electrical devices and can be distorted by the ferromagnetic materials in the environment. Beside this, the RF signals coming from the patient's body must be protected against the signals of the same frequency reaching from outside the system. The installation should have the minimum requirements to fulfill a satisfactory installation of magnetic field and RF-waves from the environment.

For this reason a special site planning should be done as explained in Chapter 7. As we saw there, two choices are possible;renovation of an old building or building a new and special one. The latter solution is better suited to the requirements of MR systems, but in most cases it costs more than a renovation, and the distance from the radiology depatment may not be practicle.

Therefore, it seems that a renovation in the present building inside the radiology department may be a better choice for a turkish public clinic, if the cost has to be kept low.

The size of the renovation depends on the type and strength of the choosen magnet. For a system with a superconducting magnet of 1.0 T, which was mentioned as the ideal magnet strength, the renovation costs decrease drastically if it is ordered to the delivering firm as equipped with "self-shielding"(see Section 7.2.1.4). In this case there is only the necessity for RF shielding. Additionally some architectural changes should be done, like building new separations in the operation room.

In our calculations we will assume that the MR system is to be installed on the basement. On the other floors, it can cause some problems because of the huge weight, which is more than 16 tonnes.

9.1.5. Medical Staff

The medical staff who is going to use the Magnetic Resonance System needs a special training. Evidently, the most suited medical doctors for the operation of an MR are the radiologists.

In order to obtain the optimal image; that is the best image in the shortest possible time, the user controlling the examination must have a very good knowledge of all the measurement and reconstruction techniques, since there is not only one method in MRI that leads to the best result, but a combination of different techniques can be made.

It must be kept in mind that the medical team using MR should comprise at least two radiologists, a physicist and two nurses. The physicist must always be ready to supervise the medical staff in technical points. The nurses are needed to help the patients before and during the examination. It is not very easy for the patients to lay down in the approximately two meter long tunnel.

The education of the medical team is necessary. The nurses, the doctors and the physicist should be

trained in an MR center for at least one month. This must be demanded from the company delivering the MR system.

9.1.6. Maintenance

The maintenance and service of an MR system is surely not very easy. Long down-times of such an expensive system is not to be efforted. That means, MR must be maintained very well and if the system is out of order, it has to be repaired very quickly.

This can be achieved only with a cooperation of the technical service of the delivering company and the biomedical engineering department of the medical institution, if there is any, which is unfortunately not always the case.

The maintenance of the MR systems will not create any problem if the following measures are taken. A biomedical engineer or at least a biomedical technician, should be trained for MR and additionally a service and maintenance contract should be signed with the delivering company.

9.1.7. Financing

According to the calculations of Chapter 8 there are four kinds of costs belonging to an MR system: capital costs, fixed and variable operating costs, and indirect costs. Table 1 and 2 in that chapter show

these expenses. In this chapter, all the expenses and other financial aspects will be evaluated for a possible MR system in Turkey, assuming that a 1.0 T device will be purchased. Costs due to interest will not be introduced into the calculations.

9.1.7.1. Capital expenses

These expenses consist of the device price, the RF shielding and renovations. We make the calculation for a 1.0 T MR system with self-shielded superconducting magnet. The purchase price of such a system is approximately US $ 1.400.000.,where self shielding costs of $ 100.000 is included. RF shielding of the room needs another $ 100.000(86). The renovation expenses are not very huge, if the installation will be realized as mentioned in Section 9.1.3. Necessary work is only some architectural changes. For such a renovation not more than TL 30.000.000 is required(x). If we take the exchange rate of $ to TL as 1:500, then we get as capital expenses the following:

$$\begin{aligned} \text{Cap.exp.} &= \$\ 1.500.000 \times 500 + \text{TL}\ 30.000.000 \\ &= \text{TL}\ 750.000.000 + \text{TL}\ 30.000.000 \\ &= \text{TL}\ 780.000.000 \end{aligned}$$

We can assume a depreciation period of 5 years. To calculate the annual capital expense, the total capital cost will be divided by 5:

$$\text{Annual Cap.exp.} = \text{TL}\ 156.000.000.$$

(x) Information is obtained from the technical service of Siemens.

9.1.7.2. Fixed Costs

Maintenance Costs: A maintenance expense of 5% of the purchase price is calculated. The purchase price being TL 750.000.000, the maintenance cost per year is approximately TL 35.000.000.(34)

Power consumption: There is power consumption of the refrigeration system and of the cooling system. The first one is 10kW, the latter one 25kW. These both systems should operate continuously without any interruption. Therefore,

25kW x 24h/day x 365 days = 219.000 kWh
219.000 kWh x TL 40/kW = TL 8.760.000

The processing power which is consumed by the computer system, control elements and room lights is negligible with respect to magnet power. By rounding the value up to TL 9.000.000. we find the annual power consumption expense.

Personnel: For a team of two doctors, one physicist and two nurses as mentioned in Section 9.1.4., the following calculation can be made:

(2 doctors + 1 physicist) x TL 200.000/month x 12 =
= TL 7.200.000
2 nurses x TL 120.000 x 12 = TL 2.880.000

Total pers.exp. = TL 10.080.000

Rent: Since paying rent is mostly not the case for a public clinic, the rent expense may be dropped.

Insurance: Normally and unfortunately the equipments in turkishpublic hospitals are not insured. But assuming the realization of the ideal case, the insurance expenses are calculated as follows :

Insurance premium = Purchase price x 0.05
= TL 15.000.000/year

Cryogen consumption: The calculation of the cryogen consumption can be taken as follows :

Liquid helium = 3500 lt/year x TL 3000/lt (xx)
= TL 10.500.000

Liquid nitrogen = 8700 lt/year x TL 280/lt (xxx)
= TL 2.450.000

Total cryogen exp. = TL 12.950.000

Miscellaneous: For the miscellaneous costs we add another TL 4.000.000 per year.

(xx) Price for 1lt liquid helium in Germany is DM 15.
(xxx) Information from Habaş A.Ş.

9.1.7.3. Variable Costs

Variable costs include the cost of film, medical supplies, paper products, magnetic disks, tapes etc. The value for this kind of cost is taken from Table 2, Chapter 8 directly, since most of these materials are imported to Turkey. We get an annual total cost of approximately TL 25.000.000.

9.1.7.4. Indirect Costs

For our case, indirect costs refer to the expenses belonging to the support from other departments of the medical institution to the MR department. Here, we can take indirect costs as 25% of the fixed costs.

So,

$$\begin{aligned} \text{Indirect costs} &= \text{fixed costs} \times 0.25 \\ &= \text{TL } 86.030.000 \times 0.25 \\ &= \text{TL } 21.507.500 \end{aligned}$$

In Table 1 the results of the calculations are given.

Table 1 Annual cost of a 1.0 T superconducting MR system (TL)

Capital Costs	
Purchase price	700.000.000
RF shielding	50.000.000
Renovation	30.000.000
	780.000.000 /5years
	= 156.000.000/year
Fixed Costs	
Maintenance	35.000.000
Power consumption	9.000.000
Cryogen consumption	12.950.000
Personnel	10.080.000
Rent	--
Insurance	15.000.000
Miscellaneous	4.000.000
	86.030.000
Variable Costs	25.000.000
Indirect Costs	21.507.500
TOTAL ANNUAL COST	288.537.500

9.1.7.5. Break-Even Analysis

It is necessary to make a break-even analysis, the point where the system recovers all its costs and becomes feasible. Today the average time necessary for a complete examination is 45 minutes. This changes according to the imaged part of the body, the head being the part with the shortest time and extremities with the longest one.

Assuming that the clinic works from 8 am to 5 pm, 10 patients can be examined in one day. For 250 working days in a year, having in mind that there will be down-times, we assume that 1500 patients can be examined in a year. The break-even point is calculated as follows:

BEP = Total annual cost/patients in a year
= TL 288.537.500 / 1500 patients
= TL 192.358 / patient

This is the minimum charge which must be paid for an examination so that the clinic can recover the costs.

9.2. Conclusion

The break-even charge of approximately TL 200.000 seems to be rather high. It can be paid only by a small portion of the population in Turkey.

A public hospital or a university clinic can effort charges under this value if it is supported by the government. Such a support to a public institution can be made in order to have up to date medical equipments in Turkey.

A reduction in the examination charge is also possible if the patient throughput or depreciation time or both increse. For example, for 3000 patients in a year and a depreciation time of 15 years, the break even charge decreses to approximately TL 65.000.

10. CONCLUSION

Magnetic Resonance is a very new diagnostic imaging method which started being used in medicine in the last few years.

MR has many advantages over other imaging techniques, including computed tomography, nuclear scanning, conventional X-ray and ultrasonography due to pictures with very high resolution, imaging the patient from all planes -coronal, sagittal, and transaxial- directly, having the possibility of differentiation between benign and malignant tumors are only some of these advantages.

The fact that there haven't appeared any side effect yet is also a great advantage of MR, since most of the other imaging methods use hazardous materials like radionuclide elements and ionizing radiation.

The benefits of MR are widely accepted by today's medical community. This acceptance will become stronger when the measurement techniques, including phosphorus-, sodium-, or fluorine imaging, chemical shift, and spectroscopy become available for routine diagnosis.

Some points which look as potential problems, including the installation and economical feasibility may be overcome. There are easy solutions for the installation difficulties as mentioned in this thesis. To cover the expenses of an MR system is less of a problem with the newly developed software packages

which allow more than 10 examinations per day.

We can conclude here that the Magnetic Resonance can be considered as one of the most important diagnostic imaging methods of today and of the near future.

For a possible installation in Turkey there are some prerequsits which must be fulfilled. For example, the medical institution should train the medical and technical staff who will operate the system. 24 hours maintenance and service must be available, so that down-times are minimized. Precise and careful work at the installation stage and precise calculation of the break-even point with realistic values are some of these prerequsits.

Additionally, the medical institution which may purchase an MR system should be a great and well organized one, having good experience with a Computed Tomography system. Magnetic Resonance is not a simple equipment that can be used like a conventional X-ray examination device. It must be operated very carefully.

APPENDIX A

GLOSSARY OF MR TERMS

Angular Momentum : A vector quantity given by the vector product of the momentum of a particle and its position vector.

Artifacts : False features in the image produced by the imaging process.

B_0 : A conventional symbol for the constant magnetic field in an MR system.

B_1 : A conventional symbol for the radiofrequency magnetic induction field used in an MR system.

Bloch equations : Phenomenological classical equations of motion for the macroscopic magnetization vector. They include the effects of precession about the magnetic field and tne T1 and T2 relaxation times.

Boltzman distribution : If a system of particles which are able to exchange energy in collisions is in thermal equilibrium, then the relative number of particles, N1 and N2, in two particular energy states with corresponding energies, E1 and E2, is given by

$$N1/N2 = \exp(-(E1-E2)/kT)$$

where k is Boltzmann's constant and T is absolute temperature.

Chemical shift : The change in the Larmor frequency of a given nucleus when bound in different sites in a molecule, due to the magnetic shielding effects of the electron orbitals.

Cryostat : An apparatus for maintaining a constant low temperature(as by means of liquid helium)

Back projection : Mathematical technique used in reconstruction fromprojections to create images from a set of multiple projection profiles. It essentially involves correcting the projection profiles by convolving them with a suitable mathematical filter and then back projecting the filtered projections into image space.

Fourier Transform : A mathematical procedure to seperate out the frequency components of a signal from its amplitudes as a function of time, or vice versa.

Gauss(G) : A unit of magnetic flux density in the older CGS system.(1T = 10.000G)

Gradient : The amount and direction of the rate of change in space of quantity, such as magnetic field strength.

Gradient magnetic field : A magnetic field which changes in strength in a certain direction.

Gyromagnetic ratio : The ratio of the magnetic moment to the angular momentum of a particle. This is a constant for a given nucleus.

Image acquisition time : Time required to carry out an MR imaging procedure comprising only the data acquisition time. The additional image reconstruction time will also be important to determine how quickly the image can be viewed.

Inversion : A nonequilibrium state in which the macroscopic magnetization vector is oriented opposite to the magnetic field.

Larmor equation : It states that the frequency of the precession of the nuclear magnetic moment is proportional to the magnetic field.

$$\omega_0 = - B_0$$

$$\text{or } f_0 = - B_0/2\pi$$

Larmor frequency : The frequency at which magnetic resonance can be excited; given by the Larmor equation.

Magnetization M : Magnetic polarization of a material produced by a magnetic field.

Magnetic moment : A measure of the net magnetic properties of an object or particle. A nucleus with an intrinsic spin will have an associated magnetic dipole moment, so that it will interact with a magnetic field.

MR signal : Electromagnetic signal in the radiofrequency range produced by the precession of the transverse magnetization of the spins.

Precession : Comparatively slow gyration of the axis of a spinning body so as to trace out a cone;

caused by the application of a torque tending to change the direction of the rotation axis, and continuously directed at right angles to the plane of the torque.

Quenching : Loss of superconductivity of the current carrying coil that may occur unexpectedly in a superconducting magnet.

Saddle coil : RF coil configuration design commonly used when the static magnetic field is coaxial with the axis of the coil along the long axis of the body.

Saturation : A nonequilibrium state in MR, in which equal numbers of spins are aligned against and with the magnetic field, so that there is no net magnetization. Can be produced by repeatedly supplying RF pulses at the Larmor frequency with interpulse times short compared to T1.

Sequential line imaging : MR imaging technique in which the image is built up from successive lines through the object.

Sequential plane imaging : MR imaging technique in which the image of an object is built up from successive planes in the object.

Sequential point imaging : MR imaging technique in which the image is built up from successive point positions in the object.

Shim coils : Coils carrying a relatively small current that are used to provide auxiliary magnetic fields in order to compensate for inhomogeneities in the main magnetic field of an MR system.

Spin density map : Figure showing the density of resonating spins in a given region.

Tesla(T) : The SI unit of magnetic flux density. One Tesla is equal to 10.000 Gauss.

Tuning : Process of adjusting the resonant frequency, e.g., of the RF circuit, to a desired value, e.g., the Larmor frequency. More generally, the process of adjusting the components of the spectrometer for optimal MR signal strength.

Voxel : Volume element; the element of 3-D space corresponding to a pixel, for a given slice thickness.

X : Dimension in the stationary frame of reference in the plane orthogonal to the direction of the static magnetic field B , z, and orthogonal to y, the other dimension in this plane.

Y : Dimension in the stationary frame of reference in the plane orthogonal to the direction of the static magnetic field B , z, and orthogonal to x.

Z : Direction in the direction of the static magnetic field B , in both the stationary and rotating frames of reference.

APPENDIX B

GLOSSARY OF MEDICAL TERMS

Abscess : Microbes, leukocytes, and liquified tissue debris walled off by fibroplasts and collagen.

Aneuryms : Abnormal widening of a blood vessel.

Axial plane : see figure appdx.1

Annulus fibrosis : Outer section of the intervertebral disks.

Basal ganglia : Several nuclei in the cerebral hemispheres which code and relay information associated with the control of muscle movements.

Brain stem : The stalk of the brain through which pass all the nerve fibers between the spinal cord and higher brain centers.

Cerebellum : One of the divisions of the brain which is involved with skeletal muscle functions.

Coronal plane : see figure appdx.1

Demyelinating diseases : Inflammation of the spinal cord.

Dural sac : Membrane in the brain

Foramen Magnum : The hole at the bottom of the skull allowing the passage of the spinal cord.

Hemorrhage : Bleeding

Hilus of lungs : The place at the bottom of lungs where the blood vessels leave the lungs.

Meningitis : Inflammation of any or all of the membranes enclosing the brain and spinal cord.

Myelin : Insulating material covering the axons of many neurons

Neoplasm : Newly growing cell, mostly tumor.

Nucleus pulposus : Inner section of the intervertebral disks.

Pituitary gland : An endocrine gland which lies in a pocket of bone just below the hypothalamus.

Sagittal plane : see figure appdx.1

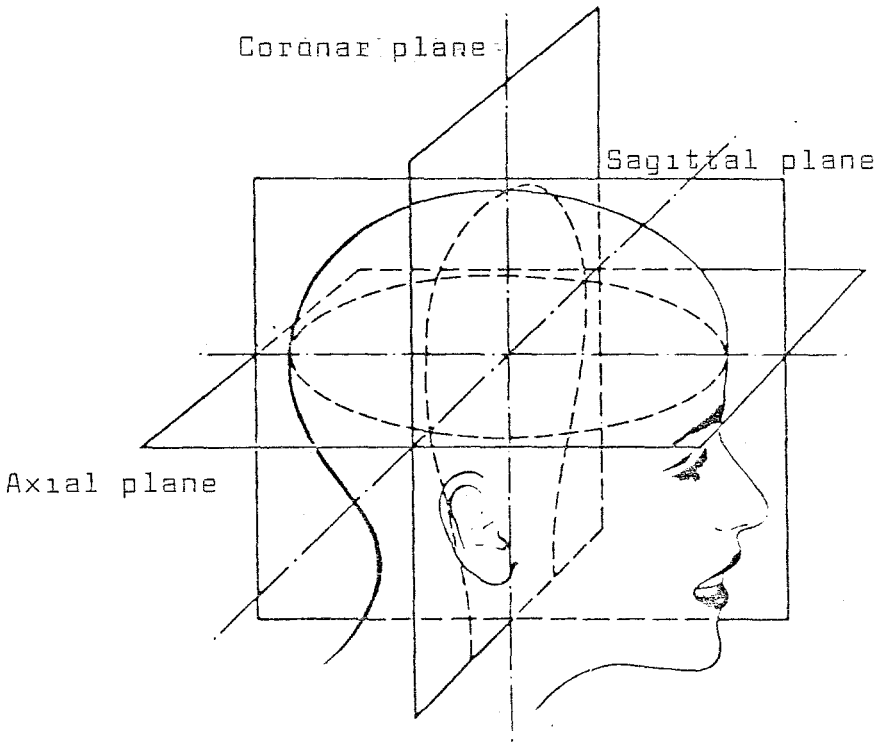
Sciatica : Neuralgia of the nerve extending throgh the hip.

Sella turcica : The bony structure housing the hypophysis.

Subarachnoid space : Space between brain membranes.

Wilson's disease : A kind of cirrhosis.

Fig. APPDX.1 Axial,coronal and sagittal planes



APPENDIX C

LIST OF ABBREVIATIONS

CT : Computed Tomography

FID : Free Induction Decay

FT : Fourier Transform

IR : Inversion Recovery

MR : Magnetic Resonance

MRI : Magnetic Resonance Imaging

PET : Positron Emission Tomography

RF : Radiofrequency

SE : Spin Echo

S/N : Signal to Noise ratio

SR : Saturation Recovery

T : Tesla

2D : Two dimensional

3D : Three dimensional

APPENDIX D

MR INSTALLATIONS

Installed MR systems in USA and in other countries are listed below. The stand is November 1984.

BRUKER
Two operating sites in the U.S. and two abroad.

Baylor College of Medicine
Dallas, TX
0.13 T resistive

Brigham & Women's Hospital
Boston, MA
0.13 T resistive
1.9 T superconducting

International sites

Japan
0.13 T resistive

D.K.D. Hospital
Wiesbaden, West Germany
0.13 T resistive

DIASONICS
Eighteen operating sites in the U.S. and four abroad. All systems are 0.5 T superconducting magnets operating at 0.35 T (except where noted)

Huntington Medical
Research Institutes
18 Pico St.
Pasadena, CA 91105
213/578-8605
Wm. Bradley, M.D., Ph.D.

UC Medical Center
San Francisco, CA 94143
415/666-4742
Charles B. Higgins, M.D.

UCSF Radiologic Imaging
Laboratory
400 Grandview Dr.
So. San Francisco, CA 94080
415/952-1374
Leon Kaufman, Ph.D.
2 T superconducting

Montclair Radiology
445 Bloomfield Ave.
Caldwell, NJ 07006
201/228-5330
Charles A. Whelan, M.D.

University of Texas HSC
5323 Harry Hines Blvd.
Dallas, TX 75235
214/637-8445
Robert Parkey, M.D.
0.35 T superconducting
2 T superconducting

U. Michigan Hospitals
Ann Arbor, MI 48109
313/763-5650
William Martel, M.D.

NMR Imaging Inc.
1930 Old Tustin Ave.
Santa Ana, CA 92701
213/318-4740
Alexander Metherell, M.D.

NMR Imaging
Torrance, CA

NMR Associates-Kirby
8968 Kirby Dr.
Houston, TX 77054
713/665-4823
Arnold Goldman, M.D.

Long Island MRI
1575 Hillside Ave.
New Hyde Park, NY 11040
516/354-4200
Dennis Rossi, M.D.

Private radiology clinic
New Jersey

Northeast Medical Center
3122 E. Commercial Blvd.
Ft. Lauderdale, FL 33308
305/772-8000
Robert L. Kagan, M.D.

Magnetic Resonance Center
6699 Alvarado Rd.
San Diego, CA 92120
619/583-6551
Jeffrey Rush, M.D.

St. Anthony's Professional
Building
1201 Fifth Ave., Suite 104
St. Petersburg, FL 33705
813/894-0181
Rex Orr, M.D.

San Jose MRI Center
2100 Forest Ave., Ste. 111
San Jose, CA 95128
408/286-5393
Cesar Mayo, M.D.

Magnetic Resonance Imagers
6449 38th Ave. North
St. Petersburg, FL 33710
813/384-0208
Allan Katz, M.D.

Heart to Heart Medical Lab
1916 W. Bethany Home Rd.
Phoenix, AZ 85015
602/249-0212
Robert Lewis, M.D.

Harris Hospital-Methodist
1300 W. Cannon Street
Fort Worth, TX 76104
817/334-6331
Richard Pickering, M.D.

International sites

Roentgeninstitut (Stork)
Kaiserwerderstrasse 89-95
D-4000 Dusseldorf 1,
West Germany
011-49-211-491038

Dr. Med. Heinz Schloesser
Schwarenstrasse 28
5600 Wuppertal-Elberfeld
West Germany
011-49-202-44-8108

Centre D'Imagene
Diagnostique
Dr. F. Azam
Grand-Chene 8
Ch-1003 Lausanne
Switzerland
011-41-21-202232

Institut De Radiologie
Rue Jean Violette 5
Ch-1205 Geneva
Switzerland
011-41-22-296510

ELSCINT
Two operating sites in the U.S. and two abroad, both systems have 0.5 T superconducting magnets.

Skokie Valley Community Hospital
Skokie, IL
0.5 T superconducting

Fondren Imaging
Houston, TX
0.5 T superconducting

International sites

Hertzelyan MRI Clinic
Herzlya, Israel
0.5 T superconducting

Private Clinic
Freiburg, West Germany
0.5 T superconducting

FONAR
Five operating sites in the U.S. and three abroad. Seven additional systems are expected to be installed by year's end.

NMR Systems Inc.

Brunswick Memorial Hospital
Amityville, NY 11701
Douglas Stein, M.D.
0.3 T permanent

UCLA
Medical Center
Los Angeles, CA 90024
William Hamilton, M.D.
0.3 T permanent

Temple University
Philadelphia, PA
Robert Liebeskind, M.D.
0.3 T mobile hybrid magnet
(owned by Universal NMR, Plantation, FL)

Montvale Diagnostic Imaging
Center
Montvale, NJ 07045
Stanley Malkin, M.D.
0.3 T permanent

Chicago Medical School
North Chicago, IL
Walid Hindo, M.D.
0.3 T permanent

Pending U.S. installations

Parkview Hospital
Hospital Corp. of America
230 25th Ave., N.
Nashville, TN 37203
David Driggs, M.D.
0.3 T permanent

Neurodiagnostic Center
New York, NY
David Godden, M.D.
0.3 T permanent

NMR Centers Inc.
Los Angeles, CA
0.3 T permanent

Mercy Hospital

Altoona, PA
(awaiting CON approval)
0.3 T permanent

Advanced Medical Diagnostics
Melbourne, FL
0.3 T permanent

NMR Investors Inc.
Santa Monica, CA
0.3 T permanent

Odessa Diagnostic Imaging
Center
Odessa, TX
0.3 T permanent

International sites

Hospital Universitario
Monterrey Nuevo Leon, Mexico
Dr. Louis Todd
0.04 T permanent

Nakatsugawa Hospital
University of Nagoya
Japan
Dr. Furusi
0.04 T permanent

San Raffaele Hospital
Milan, Italy
Dr. Giancarlo Gasparini
0.04 T permanent

GENERAL ELECTRIC
Three operating sites in the U.S. and one abroad. Nine systems are expected to be installed by year's end.

Hospital of University of
Pennsylvania
3400 Spruce St.
Philadelphia, PA 19104
Leon Axel, M.D., Ph.D.
0.12 T resistive
1.5 T superconducting

Yale-New Haven Hospital
333 Cedar St.
New Haven, CT 06510
Alexander Gorkschalk, M.D.
0.15 T resistive

Duke University
Box 3808
Durham, NC 27710
Burton Drayer, M.D.
1.5 T superconducting

Pending U.S. installations

Pittsburgh NMR Institute
130 Lothrop St., Room 145
Pittsburgh, PA 15261
1.5 T superconducting

Stanford University
Palo Alto, CA
1.5 T superconducting

Birmingham Radiological Group
Birmingham, AL
1.5 T superconducting

Committee on Emerging
Technology
Rochester, NY
1.5 T superconducting

Geisinger Clinic
Danville, PA
1.5 T superconducting

Magnetic Resonance Center
Associates
San Francisco, CA
1.5 T superconducting

Medical Imaging Centers of
America
Long Beach, CA
1.5 T superconducting

St. Mary's Hospital
West Palm Beach, FL
1.5 T superconducting

International sites

Hospitaux de Paris
Paris, France
1.5 T superconducting

M&D TECHNOLOGY
Three operating sites abroad

University of Aberdeen
Scotland, U.K.
0.04 and 0.08 T resistive

Royal Infirmary
Edinburgh, U.K.
0.08 T resistive

Private clinic
Geneva, Switzerland
0.08 T resistive

PHILIPS
Four operating sites in the U.S. and four abroad. Nine pending in installations in U.S. and abroad.

Neurological Institute of
Columbia University
Columbia-Presbyterian Med.
Center
710 West 168th St.
New York, NY 10032
Sadek Hilal, M.D.
1.5 T superconducting

Emory University
Atlanta, GA
1.5 T superconducting

New York University
Bellevue Hospital
New York, NY

Hospital Corp of America
Unidentified 250-bed hospital
0.5 T superconducting

Pending U.S. installations

University of Washington
Seattle, WA
1.5 T superconducting

U. of Southern California
Los Angeles
1.5 T superconducting

Northwestern University Hospital
Chicago
1.5 T superconducting

International sites

University of Leiden
The Netherlands
0.15 T resistive

Akademisch Ziekenhuis Leiden
Leiden, The Netherlands
0.15 T resistive
0.5 T superconducting (pending)

Casa di cura "Pio X"
Milano, Italy
0.15 T resistive
0.5 T superconducting (pending)

Instituto Neurotraumatologico
Italiano
Rome, Italy
0.15 T resistive

Pending international sites

Universita di Firenze
Florence, Italy
0.5 T superconducting

Erasmus Ziekenhuis/Free
University
Brussels, Belgium
1.5 T superconducting

Universitaetsklinik Koeln Cologne,
West Germany
0.5 T superconducting

Neuro Besta
Milano, Italy
0.15 T resistive

Montreal Neurological Institute
Montreal, Quebec
1.5 T superconducting

Centro Diagnostico Immagini
Computerizzate
Catania, Italy
0.5 T superconducting

PHOENIX INTERNATIONAL
Eleven operating sites in the U.S. and 17 abroad. Six systems are expected to be installed by year's end.

Bowman-Gray Medical School
Wake Forest University
300 S. Hawthorne Rd.
Winston-Salem, NC 27103
Charles D. Maynard, M.D.
0.15 T resistive

Mayo Clinic
200 First Street
Rochester, MN 55901
Hiller L. Baker, Jr., M.D.
0.15 T resistive

Mount Sinai Medical Center
1800 E. 105th St.
Cleveland, OH 44106
Stephen N. Wiener, M.D.
216/421-5903
0.15 T resistive

Duarte CT
1863 Business Center Dr.
Duarte, CA 91010
Charles A. Cervantes
818/357-3265
0.5 T superconducting

City of Faith Hospital
8181 Lewis Ave.
Tulsa, OK 74136
Patrick Lester, M.D.
0.5 T superconducting magnet operating at 0.3 T

University of Iowa
MRI Center
Iowa City, IA 52242
Val Dunn, M.D.
0.5 T superconducting magnet operating at 0.3 T

First Hill Diagnostic Center
1001 Boylston Ave.,
Seattle, WA 98104
Lawrence Cromwell, M.D.
0.15 T resistive

National Institutes of Health
9000 Rockville Pike
Bethesda, MD 20205
David I. Hoult, M.D.
0.5 T superconducting magnet operating at 0.3 T

Magnetic Imaging of Washington
5550 Friendship Blvd.
Chevy Chase, MD 20815

Charles M. Citrin, M.D.
0.5 T superconducting
Elkins Park
Philadelphia
0.15 T resistive

Imaging Systems Inc.
Bellevue, WA
0.5 T superconducting

Pending U.S. installations

University of Alabama
933 10th Ave., S.
Birmingham, AL 35294
Jerry Gibbon, Ph.D.
0.5 T superconducting magnet to be installed

Diablo Valley Radiology
Walnut Creek, CA
0.5 T superconducting

Holy Family Hospital
Spokane, WA
0.5 T superconducting

The Neurology Center
Pennsylvania
0.5 T superconducting

NMR Scanning Laboratories
Oak Brook, IL
0.5 T superconducting

Outpatient Radiology
Portland, OR
0.5 T superconducting

International sites

University of British Columbia
Vancouver
0.35 T superconducting

Shinsuma Hospital
Kobe University
Kobe, Japan
0.15 T resistive

Riyadh, Saudi Arabia
0.15 T resistive

Chiba University
Chiba City, Japan
0.3 T superconducting

National Heart Institute
London
0.5 T superconducting

Singapore
0.3 T superconducting

Hammersmith Hospital
London
0.3 T superconducting magnet operating at 0.15 T

Queens Medical Center
Nottingham, England
0.1 T resistive

Nottingham University
Nottingham, England
0.12 T resistive

HIRST Research Center
London
0.15 T resistive

Private clinic
500 Koln 80
Cologne, Germany
Dr. Assheuer
0.3 T superconducting

Private clinic
6500 Mainz 1
Munich, Germany
Dr. Wallnhofer
0.3 T superconducting

Cologne, Germany
0.5 T superconducting

Frankfurt, Germany
0.5 T superconducting

University of Manchester
England
0.3 T superconducting

Queens Square Hospital
London
0.3 T superconducting

Glasgow, Scotland
0.15 T resistive

[illegible]
Seven operating sites in the U.S. and eight abroad. Nineteen systems expected to be installed by year's end.

Mallinckrodt Institute
510 South Kingshighway
St. Louis, MO 63110
0.5 T superconducting magnet operating at 0.35 T
1.5 T system to be installed
Ronald G. Evens, M.D.

St. Vincent Medical Center
2131 W. Third St.
Los Angeles 90057
Victor Waluck, M.D.
1 T superconducting magnet operating at 0.35 T

Allegheny General
320 E. North Ave.
Pittsburgh, PA 15212
Rolf L. Schapiro, M.D.
0.5 T superconducting magnet operating at 0.35 T

Mount Sinai
Medical Center
Miami Beach
Manual Viamonte, Jr., M.D.
0.5 T superconducting magnet operating at 0.35 T

Columbia, MO
Vijay K. Sachu, M.D.
0.5 T superconducting magnet operating at 0.35 T

Memorial Hospital Health Center
Box 1428
Long Beach, CA 90801
William J. Wilson, M.D.
1 T superconducting magnet operating at 0.35 T

Radiology Associates
Little Rock, AK
H. Howard Cockrill, Jr., M.D.
0.5 T superconducting magnet operating at 0.35 T

Pending U.S. installations

Pomona Valley Community Hospital
1798 N. Garey Ave.
Pomona, CA 91767
0.5 T superconducting magnet operating at 0.35 T

Loma Linda U. Medical Center
11234 Anderson St.
Box 2000
Loma Linda, CA 92354
0.5 T superconducting magnet operating at 0.35 T

Hershey Medical Center
500 University Dr.
Hershey, PA 17033
William A. Wedner, M.D.
717/534-8044
2 T superconducting

St. Francis Medical Center
Peoria, IL
Robert M. Wright, M.D.
0.5 T superconducting magnet operating at 0.35 T

Digital Diagnostics
Baton Rouge, LA
Charles G. Greeson, M.D.
0.5 T superconducting magnet operating at 0.35 T

Magnetic Resonance Imaging Center
Brooklyn, NY
0.5 T superconducting magnet operating at 0.35 T

Wendover Park Associates
Greensboro, NC
James M. Love, M.D.
0.5 T superconducting magnet operating at 0.35 T

University of Minnesota
St. Paul, MN
Eugene Gedgaudes, M.D.
0.5 T superconducting magnet operating at 0.35 T

Ochsner Foundation
New Orleans

Christopher R. Merritt, M.D.
0.5 T superconducting magnet operating at 0.35 T

University of Virginia
Charlottesville, VA
Charles D. Teates, M.D.
0.5 T superconducting magnet operating at 0.35 T

American Shared Hospital Services
San Francisco
0.5 T superconducting magnet operating at 0.35 T

Pacific Medical Centers
San Francisco
Kurt Moon, M.D.
0.5 T superconducting magnet operating at 0.35 T

Nebraska Methodist/NMR Inc
Omaha
0.5 T superconducting magnet operating at 0.35 T

NMR Imaging Associates
Marlton, NJ
Charles Goldstein, M.D.
0.5 T superconducting magnet operating at 0.35 T

Southwest Texas Methodist
San Antonio
James Stewart, M.D.
0.5 T superconducting magnet operating at 0.35 T

Long Island Jewish Hillside Medical Center
New Hyde Park, NY
0.5 T superconducting magnet operating at 0.35 T

Methodist Hospital
Houston
0.5 T superconducting magnet operating at 0.35 T

New England Medical Center
Boston
0.5 T superconducting magnet operating at 0.35 T

University Diagnostic Institute
Tampa, FL
Martin T. Sytti, M.D.
0.5 T superconducting magnet operating at 0.35 T

International sites

Private group
Munich, West Germany
Dr. Heiler
0.5 T superconducting

University of Berlin
Berlin, West Germany
0.5 T superconducting

Frankfurt, West Germany
Dr. Kuehnert
0.5 T superconducting

University of Heidelberg
Heidelberg, West Germany
0.5 T superconducting

Private group
Solingen, West Germany
0.5 T superconducting

University of Tokyo
Tokyo, Japan
0.5 T superconducting

University of Upsala
Upsala, Sweden
0.5 T superconducting

University of Hanover [Germany]
Medical Center
0.2 T resistive
0.5 T superconducting

TECHNICARE
Forty-nine operating sites in the U.S. and nine abroad. Technicare declined to release a list of its operating sites. This incomplete list was compiled from independent sources.

Scottsdale Memorial Hospital
7400 Osborn Rd.
Scottsdale, AZ 85251
0.6 T superconducting

AMC Cancer Research Center & Hospital
6401 West Colfax Ave.
Lakewood, CO 80214

Shands Teaching Hospital
University of Florida
Hospital Box J 326
Gainesville 32610
0.35 T superconducting

St. Luke's Hospital
1900 Boulevard
Jacksonville, FL 32206
Karen Matthews

904/359-3700
0.15 T resistive

Rush-Presbyterian-St. Luke's Hospital
1753 W. Congress Parkway
Chicago 60612

Greenberg Radiology Clinic
1150 Park Ave. West
Highland Park, IL 60035

Indiana University
400 W. Michigan Ave.
Indianapolis, IN 46226
0.15 T resistive

University of Kentucky
Lexington, KY 40536-0084
Harold D. Rosenbaum, M.D.
0.15 T resistive

Massachusetts General Hospital
32 Fruit St.
Boston 02114
Juan M. Taveras, M.D.
0.15 T resistive
0.6 T superconducting

Millard Fillmore Hospital
3 Gates Circle
Buffalo, NY 14209
0.15 T resistive

North Shore University Hospital
300 Community Dr.
Manhasset, NY 11030
Harry L. Stein, M.D.
516/562-4800
0.6 T superconducting

Messina & Liebeskind
926 5th Ave.
New York, NY 10021

New York Hospital
525 E. 68th Street
New York 10021
0.5 T superconducting

Charlotte Memorial Hospital
1000 Blyth Blvd.
Charlotte, NC 28204
0.15 T resistive

Cleveland Clinic Foundation
9500 Euclid Ave.
Cleveland, OH 44106
Thomas F. Meaney, M.D.
0.15 T resistive
0.6 T superconducting
1.5 T superconducting

University Hospitals of Cleveland
2074 Abington Rd.
Cleveland, OH 44106
Ralph J. Alfidi, M.D.
216/444-3858
0.3 T superconducting
1.5 T superconducting

Hershey Medical Center
500 University Dr.

Hershey, PA 17033
William A. Weidner, M.D.
717/534-8044
0.15 T resistive

Vanderbilt University
Nashville, TN 37232
C. Leon Partain, Ph.D., M.D.
0.5 T superconducting

Baylor University
3500 Gaston Ave.
Dallas, TX 75246
Steven Harms, M.D.
0.15 T superconducting
0.3 T superconducting
1.5 T superconducting

Houston Imaging Center
7000 Fannin St.
Houston 77030

Broward NMR
Fort Lauderdale, FL
0.5 T superconducting

University Park Imaging
Urbana, IL
0.6 T superconducting

VA Medical Center

St. Louis
0.15 T resistive

Temple Radiology
New Haven, CT
0.15 T resistive

Albert Einstein
Medical Center
Philadelphia
0.15 T resistive

Nuclear Facilities
Brooklyn, NY
0.5 T superconducting

NMR Diagnostic Center
Sun City, AZ
0.15 T resistive

Garden State
Medical Center
Marlton, NJ
0.6 T superconducting

Magnetic Imaging of Bellville
Bellville, IL
0.6 T superconducting

Private clinic
Union, NJ
0.6 T superconducting

Ft. Worth Magnetic
Imaging Institute
Ft. Worth, TX
0.5 T superconducting

International sites

St. Joseph's Hospital
268 Grosvenor St.
London, Ontario N6A 4V2
Canada
0.15 T resistive

Ontario Cancer Institute
500 Sherbourne St
Toronto, Canada M4X 1K9
0.15 T resistive

Hospital Hausstein
Munich, Germany
0.15 T resistive

NMR SA
Barcelona, Spain
0.15 T resistive

Clairval Hospital
Marseille, France
0.15 T resistive

Private clinic
Hanover, West Germany
0.5 T superconducting

Clinique du Park
Paris, France
0.15 T resistive

Private clinic
Antwerp, Belgium
0.15 T resistive

TOSHIBA
Five operating sites in Japan

Toshiba Central Hospital
Tokyo, Japan

Tokyo Tikei Medical College
Tokyo, Japan

Okayama University Hospital
Okayama prefecture, Japan

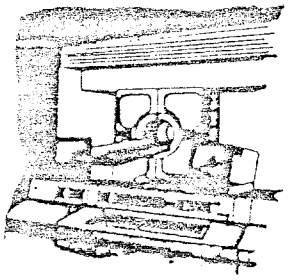
Aichi Diagnostic Technology
Committee Hospital
(assisted by Nagoya University)
Aichi prefecture, Japan

Koga Hospital
Fukuoka prefecture, Japan

APPENDIX E

MR Systems that are produced by various manufactures are listed below:

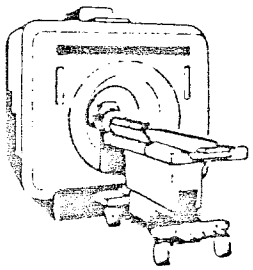
Siemens



MAGNETOM-M5 (-M10 - H15 - H20)

Magnet: Superconductive medium-field magnet 0.5 T
Version M10: medium-field magnet 1.0 T
Version H15: strong-field magnet 1.5 T
Version H20: 2.0 T strong-field magnet, charged to 1.5 T
RF-Coil 55 cm Ø fixed; tunnel aperture 69 cm Ø
Gradient field strength 3 mT/m (X,Y,Z)
RF system: body coil - measuring volume 50 cm Ø spherical
Resolution ≤ 2.4 mm
Head coil 25 cm Ø spherical, resolution ≤ 1.2 mm, optional magnet direct shielding;
Slice thickness 5 - 20 mm, test frequencies 14.9 - 21.3 - 42.6 - 63.9 MHz
Host computer: VAX 11/730, 32 bit structure
Main memory 2 MB, 2 x 25 MB fixed / moving-arm disk
121 MB disk memory; 456 MB fixed disk optional
Image processor: BSP 11/MR array processor
2 D and 3 D Fourier transformation
Measuring matrix 128 x 128; 256 x 256; interpolated 512 x 512
Measuring modality: spin echo (SE), inversion recovery (IR)
Summation mode T_1; T_2 weighted images; zoom, ECG triggering,
Respiration synchronisation
Multiple slice exposures SE max. 15 slices, multiple echo up to 32 echoes
IR max. 4 slices; 3 D operation
Patient table: table can be moved vertically and horizontally

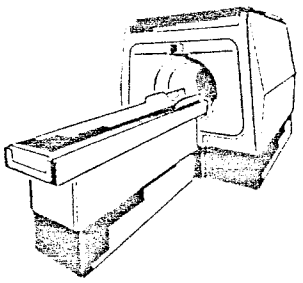
General Electric



SIGNA

Magnet: Superconductive 0.5 T, 1.0 T, 1.5 T - Oxford Magnet Technique
RF coil 55 cm Ø; tunnel aperture 55 cm - tapered
Gradient field strength up to 10 mT/m (X,Y,Z)
RF coil: body coil - measuring field 16 - 48 cm axial
16 - 40 cm coronal/sagittal
Head coil 16, 20, 24 cm in X, Y, Z-planes
Slice thickness 3; 5; 10; 15; 20 mm
Frequency 10 - 80 MHz; primary 63.9 MHz (1.5 T)
Host processor: Data General Eclipse MV 4000, 32 bit
2 MB main memory; 2 x 354 MB fixed-disk store
Image processor: Array processor, 2 D and 3 D image reconstruction
Measuring matrix 128 x 256, 256 x 256
Measuring modality: partial saturation (PS); Inversion recovery (IR)
Spin Echo (SE); multiple echo up to 4 echoes
Multiple slice exposures: SE max. 24 slices
IR max. 1 slice; PS max. 10 slices
Patient table: trolly system can be latched to magnet

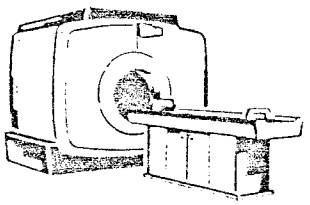
Technicare



TESLACON

Magnet: Superconductive - Oxford Magnet Technique - 0.5 T - 0.6 T - 1.5 T
Resistive - Magnet Corp. of America - 0.15 T
RF coil 53 cm Ø
RF-System: body coil 53 cm, head coil 27 cm
Resolution 1.3 mm
Host processor: PDP 11 - 24, 16 bit structure
0.5 MB main memory; 160 MB disk memory
Image processor: array processor; 2 D and 3 D Fourier transformation
Measuring matrix 128 x 128; 256 x 256, interpolated 512 x 512
Measuring modality: saturation recovery (SR); spin echo (SE)
Inversion recovery (IR); 2 D max. 64 slices (?)
3 D max. 128 slices (?)
Patient table: can be moved vertically and horizontally

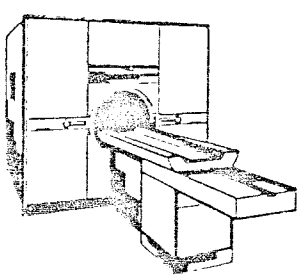
Diasonics



MR - MAGNET MT/S

Magnet: Superconductive-Oxford Magnet Technique - 0.35 T - 0.7 T
RF coil ?
RF system: 4 coils, head, neck, body circle and ellipse
Resolution - head 0.8 and 1.7 mm; body 1.4 and 2.4 mm
Slice thickness 2 mm and 5 mm
Host processor: VAX 11/730 - 32 bit - 2 MB
Image processor: array processor - Minimap
2 D Fourier transformation; Measuring matrix 128 x 128; 256 x 256
Measuring modality: spin echo (SE); inversion recovery (IR),
Partial saturation (PS), ECG triggerung, multiple echo up to 5 echoes
20 slices possible, 3 D operation
Patient table: fixed height (of wood, without brakes)

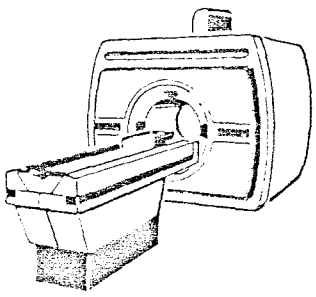
Picker



VISTA - MRI-System

Magnet: Superconductive 0.3 T - 0.5 T Oxford Magnet Technique
Resistance 0.15 T
Tunnel aperture 53 cm Ø; RF coil 53 cm Ø
RF system: body coil - measuring field 53 x 35 x 61 cm - elliptical
Head coil 30 cm Ø; slice thickness 10; 20 mm
Host computer: Perkin Elmer 3220 - 32 bit
0.75 MB main memory; 160 MB disk memory
Image computer: back-projection
Measuring matrix 256 x 256, interpolated 512 x 512
Measuring modality: inversion recovery (IR); partial saturation (PS)
Spin echo (SE); T_1; T_2 images; 16 slices possible
Patient table: can be moved horizontally, fixed vertically

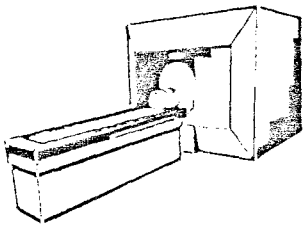
CGR



MAGNISCAN 5000

Magnet: Superconductive 0.15 T; 0.35 T; 0.5 T
RF system: head coil 30 cm Ø; body coil 55 cm Ø
Slice thickness 4 - 300 mm; reconstruction time: 5 s (256^2), 30 s (512^2)
Host processor: VAX 11/730; main memory 1 MB
Disk memory 120 MB for program
456 MB for raw data and images
Measuring modality: inversion recovery (IR); spin echo (SE)
Saturation recovery (SR); 16 slices; 2 D FT
Patient table: movable horizontally and vertically

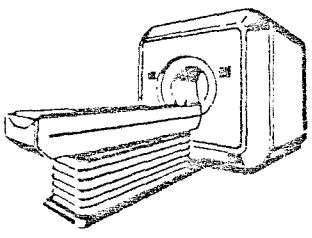
Elscint



GYRIX S 5000 / S 3500

Magnet: Superconductive 0.5 T / 0.35 T
RF system: head coil 30 cm Ø; body coil 57 cm x 37 cm - elliptical
optional 15 cm Ø coil for extremities
Slice thickness 5; 10; 20 mm - 20 slices in 5 minutes
Host processor: PDP 11/24
Image processor: array processor, Elscint Multiprocessor
Measuring modality: inversion recovery (IR); spin echo (SE)
Saturation recovery (SR); optional cardiac triggering, 2 D
Patient table: movable horizontally and vertically

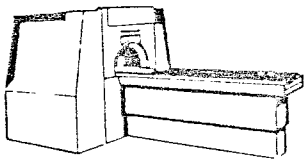
Toshiba



MRT - 50 A (- 22 A)

Magnet: Superconductive 0.5 T
Version 22 A: resistive 0.22 T

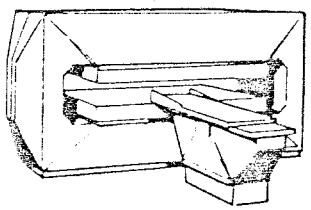
Philips



GYROSCAN S5 (- S15)

Magnet: Superconductive 0.5 T; version S15: 1.5 T
RF coil > 50 cm Ø; tunnel aperture 68 cm
Gradient field strength ± 3 mT/m (X,Y,Z)
RF system: body and head coil 29 cm to 65 cm Ø
Resolution up to 1 mm, slice thickness 2.5 mm - 100 mm
Host computer: VAX 11/750; 456 MB disk memory
Image processor: array processor; 2 D and 3 D Fourier transformation
Measuring matrix 64 x 64; 128 x 128; 256 x 256
Measuring modality: spin echo (SE); inversion recovery (IR); Partial saturation (P
Multiple slice technique possible; ECG triggerung
T_1; T_2 images; multiple echo up to 8 echoes
Patient table: horizontally and vertically movable

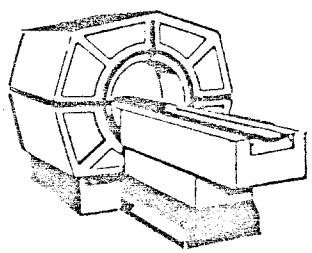
Fonar



BETA 3000 (M)

Magnet: Permanent 0.3 T; weight 20 t
RF coil 30 cm x 50 cm; tunnel aperture 30 cm x 90 cm
RF system: body coil 30 cm x 50 cm; head coil approximately 28 cm Ø
Resolution approximately 1.25 mm, slice thickness 8 mm, frequency approx. 12 MHz
Host processor: Data General Eclipse S140
0.5 MB main memory; 147 MB disk memory
Image processor: Modular Image Processor (Analogic Corp.)
Measuring matrix 256 x 256; interpolated 512 x 512
2 D Fourier transformation
Measuring modality: spin echo (SE); inversion recovery (IR)
Saturation recovery (SR); 16 slices possible
Patient table: movable horizontally and vertically
Version Beta 3000 M: for installation in a vehicle

Bruker



BNT 1000 / 1100

Magnet: Resistive 0.24 - 0.28 T
RF coil ? Ø
RF system: slice thickness 12 - 14 mm
Host processor: Aspect 3000
Image processor:
Measuring matrix 128 x 128, 256 x 256; interpolated 512 x 512
2 D Fourier transformation; back-projection
Measuring modality: spin echo (SE); up to 23 slices
T_1, T_2 images; multiple echo

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