PRAKASH NARAIN TANDON
RAVI RAMAMURTHI

Volume 1

RAMAMURTHI & TANDON'S TEXTBOOK OF NEUROSURGERY

3rd Edition

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Ramamurthi and Tandon's

Vol 1

Textbook of Neurosurgery

Third Edition

Prakash Narain Tandon
MBBS MS FRCS DSc (h.c.) FNA FNASc FASc FTWAS FAMS FRSM
Emeritus Professor of Neurosurgery
All India Institute of Medical Sciences, New Delhi
President, National Brain Research Centre Society
Manesar, Gurgaon, Haryana, India

Ravi Ramamurthi
MS FRCS Ed (SN)
Head of the Department of Neurosurgery
Dr Achanta Lakshmipathi Neurosurgical Centre
Postgraduate Institute of Neurological Surgery
Voluntary Health Services Hospital
Taramani, Chennai, Tamil Nadu, India

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10 Basic Principles of CT Scan and MRI Scan

Veena Norhona

Basics of Computed Tomography

INTRODUCTION

The first computed tomography (CT) scanner was developed by Sir Godfrey Hounsfield in 1972; since then this modality has become an important tool in diagnostic radiology. Since the first scanner to the present day multislice helical scanner, CT technology has revolutionised the world of imaging and enhanced patient management.

BASIC PHYSICS

Computed tomography uses X-rays to obtain cross-sectional, two-dimensional (2-D) images of the body. The cross-sectional image is produced by 360° rotation of the X-ray tube around the patient. The transmitted radiation is measured by the detectors located inside the gantry like a ring around the patient. The final image is generated from these measurements. The gantry of the CT machine houses the X-ray tube and the detectors (Fig. 1).

TYPES OF SCANNING TECHNIQUES

Axial (Sequential) Scanning

In sequential scanning, a single slice is obtained with a single 360° rotation of the tube (Fig. 2A). The disadvantage is that the time taken for an individual study is long, hence prone to motion artifacts and the quality of reformations is suboptimal.

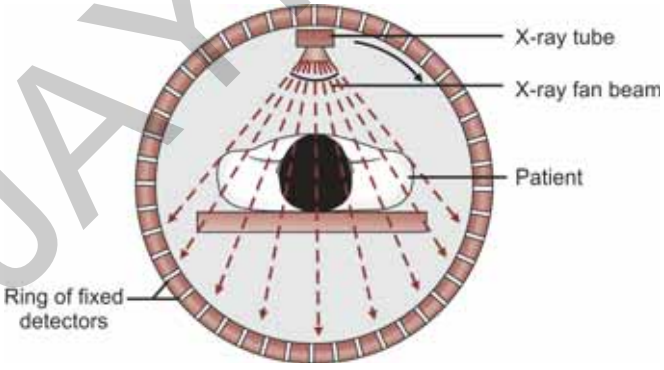


Fig. 1: Cross-sectional view of the gantry showing the orientation of the X-ray tube and detectors in a fourth generation CT scanner

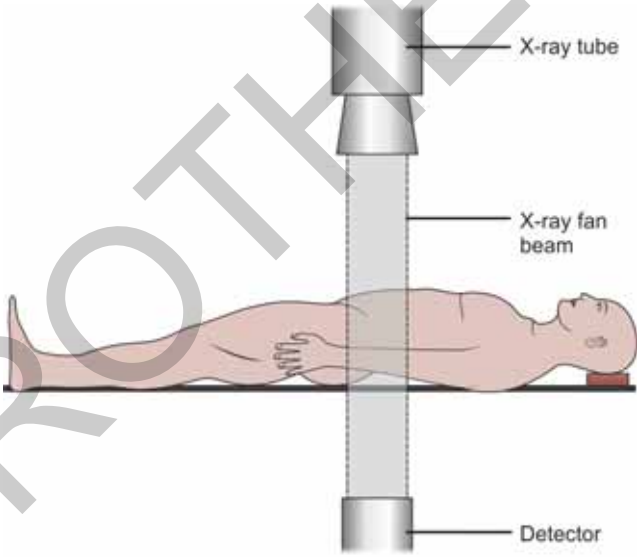


Fig. 2A: Sequential scan—single cross-sectional slice of the patient in a single rotation

Helical (Spiral) Scanning

With the advent of slip ring technology, the continuous rotation of the X-ray tube around the patient is made possible during continuous patient table movement. This led to the development of helical scanning (Fig. 2B). The transmitted radiation thus, takes the form of a helix or spiral around the patient acquiring a large volume of data. This allows larger anatomical regions of the body to be imaged during a single breath hold, thereby reducing the

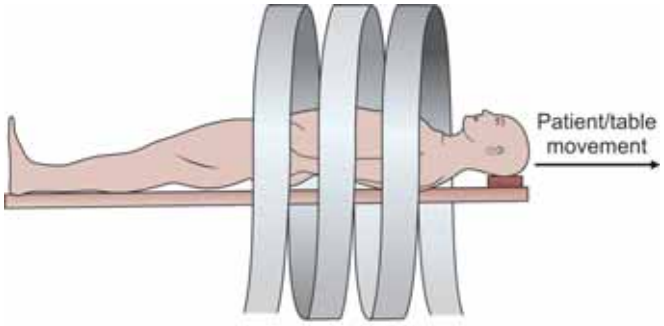


Fig. 2B: Helical scan—rotation of the tube around the patient with continuous table movement

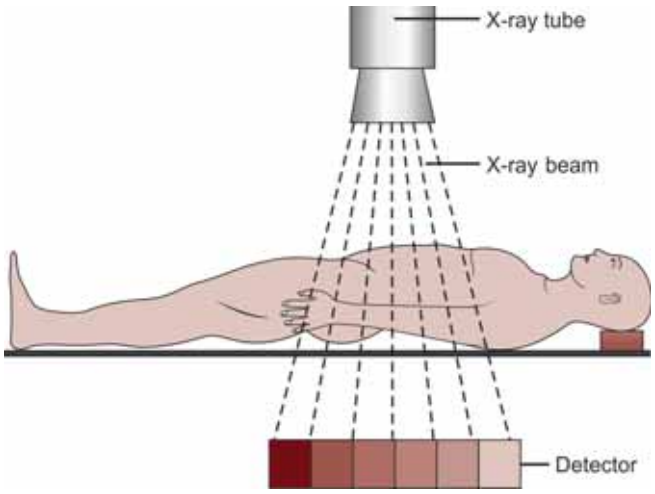


Fig. 3: Multi-slice imaging—generation of six slices per rotation of the tube in a multi-detector scanner

possibility of artifacts caused by patient movement. Faster scanning also increases patient throughput.

Multi-slice or multi-detector machines utilise the principles of the helical scanner but incorporate multiple rows of detector rings. They can therefore acquire multiple slices per tube rotation, thereby increasing the anatomical coverage in a shorter time (Fig. 3).

COMPUTED TOMOGRAPHY TERMINOLOGIES

Pixel and Voxel

Every CT image is made up of a square of picture elements called the pixel and volume element called the voxel (Fig. 4). The obtained CT image is subdivided into a matrix of up to 512 × 512 or 1024 × 1024 elements. The pixel width is determined by the field of view (FOV) and matrix size, i.e. FOV/matrix. The voxel volume = pixel area × slice thickness.

Hounsfield Unit or Computed Tomography Number

Each voxel is traversed during the scan by numerous X-ray photons and the intensity of the transmitted radiation is measured by the detectors. From these

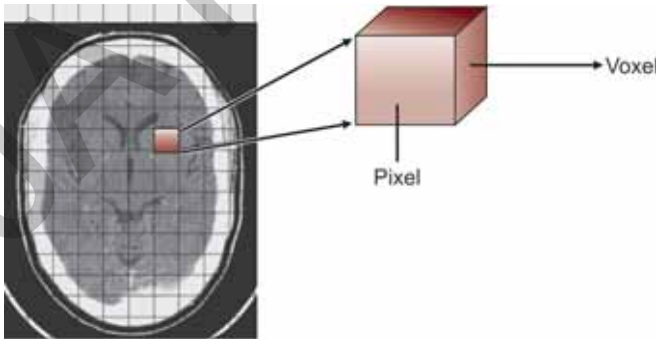


Fig. 4: Pixel—represents the matrix and voxel represents the slice thickness

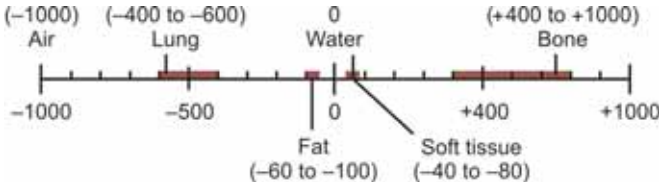


Fig. 5: Scale representing the range of Hounsfield numbers of the tissues seen in the body

intensity readings, the density or attenuation value, viz. Hounsfield unit (HU) or CT number is calculated and assigned to every tissue.

Each pixel is assigned a numerical value (CT number), based on the attenuation of X-rays by the tissue. This number is compared to the attenuation value of water and displayed on a scale of arbitrary units named HU after Sir Godfrey Hounsfield. This scale assigns water an attenuation value (HU) of zero. Each number represents a shade of grey with +1000 (white) and –1000 (black) at either end of the spectrum (Fig. 5).

The CT number of various tissues in the body is as follows:

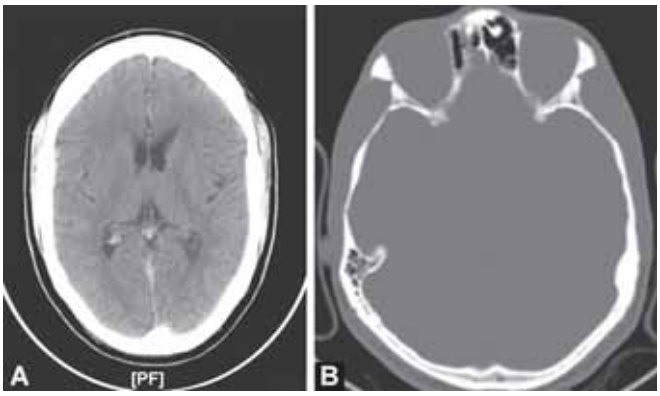
- Brain grey matter — 35–40 HU
- Brain white matter — 30–35 HU
- Blood—Flowing blood — 40 HU
- Acute haematoma — 70–90 HU (density depends on the haemoglobin concentration and coagulation profile)
- Calcification — + 80 and above
- Fat — – 100
- CSF — 0–10 HU
- Bone — + 800–1000 (depends on the type of bone).

WINDOW LEVEL AND WINDOW WIDTH

The term "window level" (WL) represents the central Hounsfield unit of all the numbers within the window width (WW). The WW covers the HU of all the tissues of interest and these are displayed as various shades of grey. Tissues with CT numbers outside this range are displayed as either black or white. Both the WL and WW can be set independently on the computer console and their respective settings affect the final displayed image (Figs 6A and B).

SLICE THICKNESS

It is the collimation of the X-ray beam as it emerges from the X-ray tube. The slice thickness can be varied depending on the anatomical region to be covered by varying the beam collimation. For example orbit scanning is done using 2–3 mm slice thickness, posterior fossa 4–5 mm slice thickness and supratentorial brain parenchyma 10 mm slice thickness.



Figs 6A and B: (A) Soft tissue window settings of an axial CT scan of the brain (WW = 100, WL = 30). (B) Bone window setting of an axial CT scan of the brain (WW = 2,000, WL = 220)

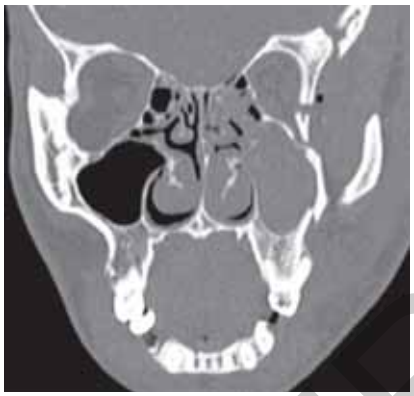


Fig. 8: Coronal reformations of the face showing fractures involving the lateral wall of the left maxillary sinus, zygoma, lateral wall of the left orbit and frontal bone

PITCH

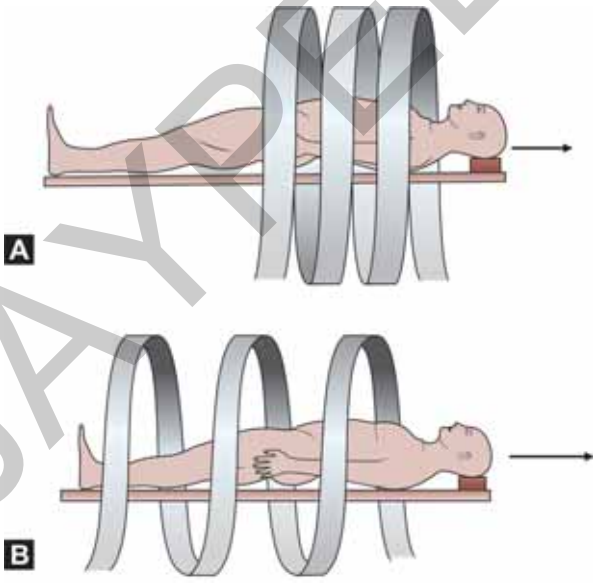
Pitch is the terminology used in helical scanning and denotes the distance travelled by the table (in millimetres) during one complete rotation of the X-ray tube, divided by the slice thickness (millimetres). Increasing the pitch by increasing the table speed reduces dose and scanning time, but at the cost of decreased image resolution (Figs 7A and B).

$$\text{Pitch} = \frac{\text{Table distance per } 360^\circ \text{ rotation (mm)}}{\text{Slice thickness (mm)}}$$

IMAGE POST PROCESSING

Post processing the acquired volumetric data during spiral CT is done in ways appropriate to the clinical situation such as:

- *Multiplanar reformatting*: After obtaining the serial axial volumetric data, the computer reconstructs the data in sagittal and coronal planes. With the current multi-slice CT scanner it is possible to obtain isotropic sagittal and coronal reconstructions. These are useful in paediatric and trauma patients who cannot be positioned for direct coronal scans (Fig. 8).
- *Three-dimensional (3-D) imaging*: The acquired data can also be post processed to obtain a 3-D model to display spatial information or surface characteristics (volume and surface rendering). This is useful in paediatric craniofacial anomalies and maxillofacial injuries to guide the surgeon in treatment planning (Fig. 9).
- *CT angiography (CTA)*: This involves injection of 100–120 ml of contrast medium, rapidly, using a pressure injector at a predetermined rate of injection. Serial axial images are obtained. These images are then used for reconstruction of the data using maximum intensity projection to get a display of the vascular tree. By altering the time of image acquisition and contrast injection, we can obtain only the arterial or venous phases (Figs 10A and B).



Figs 7A and B: (A) Shows a low pitch—tight helic. (B) A pitch of more than 1 – loose helic – shorter scan time at the cost of image resolution

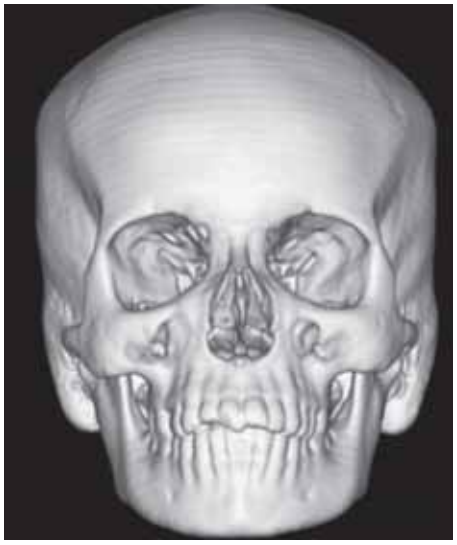
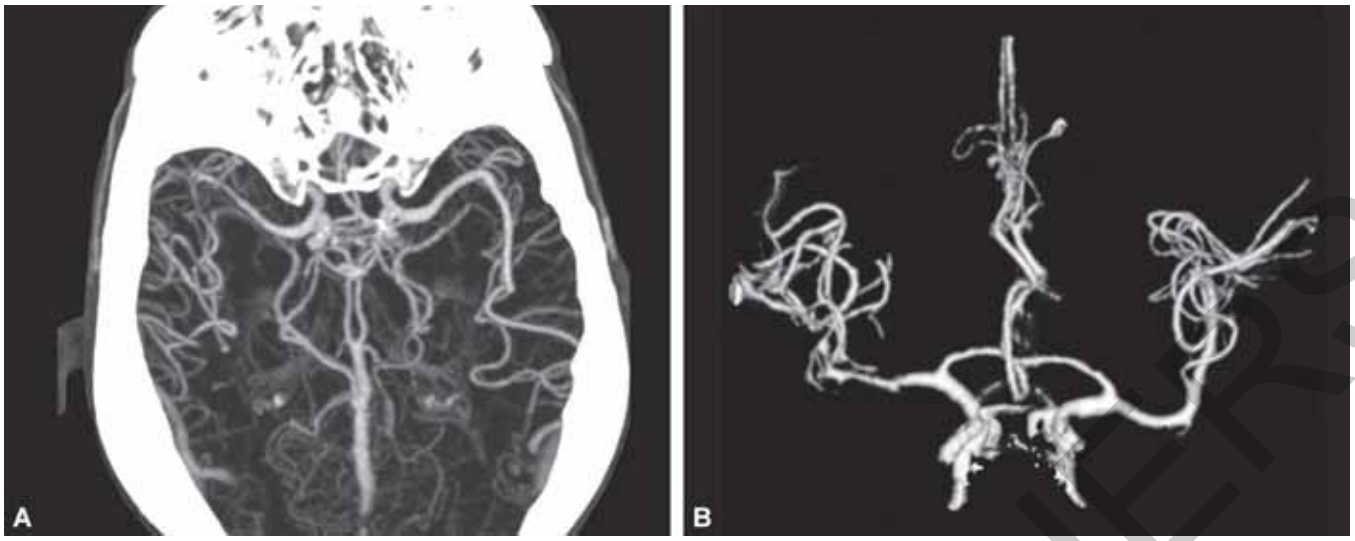


Fig. 9: CT of the skull



Figs 10A and B: CT angiogram of the brain showing. (A) Axial MIP image. (B) Volume rendering

COMPUTED TOMOGRAPHY CONTRAST MEDIA

These are iodine containing compounds. Iodine absorbs X-rays within the CT range (120 KVp) since iodine has an atomic number of 53 and atomic weight of 127.

There are two types of contrast agents used:

1. *Ionic contrast*: These are sodium or methylglucamine combined with a tri-iodinated benzene ring to form soluble salts. These are hyperosmolar and hence are likely to cause severe contrast reactions. These are contraindicated intrathecally.
2. *Non-ionic contrast*: These are near iso-osmolar and hence tend to produce fewer side effects and considered relatively safe for patients.

Absolute contraindication for contrast:

1. Previous contrast sensitivity
2. Abnormal renal parameters

Patients with diabetes and multiple myeloma are more likely to develop altered renal function post IV contrast injection. Patients with myasthenia gravis, sickle cell anaemia and pheochromocytoma are at risk of developing contrast-induced symptoms.

ADVANTAGES AND CLINICAL USE OF COMPUTED TOMOGRAPHY

- CT is readily available in most hospitals and is cost-effective.
- It is a rapid imaging modality with excellent image resolution, hence useful in trauma, paediatric and uncooperative patients.
- Patients in whom magnetic resonance imaging (MRI) is contraindicated.

DISADVANTAGES OF COMPUTED TOMOGRAPHY

- Radiation—The effective doses from diagnostic CT procedures are typically estimated to be in the range of 1–10 mSv.

FURTHER READING

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Physical Principles of Magnetic Resonance Imaging

Magnetic resonance imaging (MRI) is based on the principles of nuclear magnetic resonance (NMR).

BASIC PRINCIPLES OF MAGNETIC RESONANCE IMAGING

Magnetic resonance imaging is based on the absorption and emission of energy in the radiofrequency (RF) range of the electromagnetic spectrum. The human body is primarily made up of fat and water, which have many hydrogen atoms (almost 63%) (Fig. 11). The hydrogen atom (1H) consists of a single positively charged proton that spins around its axis. These charged particles create an electromagnetic field, similar to that of a bar magnet.

The proton possesses a property, called spin, which has a small magnetic field. These spinning particles have a net magnetic moment which has both magnitude and direction. In the absence of an external magnetic field, these protons are randomly oriented.

When placed in a magnetic field of strength B, the protons align themselves parallel or antiparallel to the external magnetic field. There is a low energy state where the poles are aligned N-S-N-S and a high energy state N-N-S-S. This particle can undergo a transition between the two energy states by the absorption of a photon. A particle in the lower energy state absorbs a photon and ends up in the upper energy state. The energy of this photon must exactly match the energy difference between the two states.

Application of a RF pulse of appropriate duration and amplitude excites these protons from the lower energy state to the higher energy state.

The MRI signal results from the energy difference of the spins emitted during transition from the higher energy state to the lower energy state. The signal is thus proportional to the population difference between the states (Figs 12A and B).

When the RF pulse is applied, the protons are tipped into the horizontal or X-Y plane by an angle termed as the flip angle or tip angle depending on the type of RF pulse. The rate at which the protons precess is termed as frequency and the angular position of the precessing spin is called the phase of the spin.

The frequency of precession (f) is called the Larmor frequency and is characteristic of the specific nucleus and strength of the external magnetic field and is expressed as:

$$f = \gamma B$$

Where f = mHz/sec, B is expressed in Tesla and ã is the gyromagnetic ratio of the specific nucleus and expressed as mHz/T. Hydrogen has the highest gyromagnetic ratio and is the most abundant body element, hence is the natural choice for H signal.

Radiofrequency Field

Every nucleus in the body precesses at its own Larmor frequency and will produce an MR signal only when the RF energy is delivered at the correct frequency. The excitation RF pulses are delivered by coils that produce an RF field perpendicular to the external magnetic field. The RF is absorbed by the nuclei and the magnetic moment is tipped away from the Z axis, i.e. axis of the external magnetic field depending on the duration and amplitude of the RF pulse.

Free Induction Decay

When the RF pulse is switched off, the magnetic momentum of the nuclei begins to return to its original position, thereby transferring the absorbed energy and inducing alternating current signal in the receiver coil. This is termed as free induction decay (FID). As this occurs immediately after the RF pulse, this signal is not used for image data. The magnetisation is manipulated to generate a useful signal termed as echo, which produces the image.

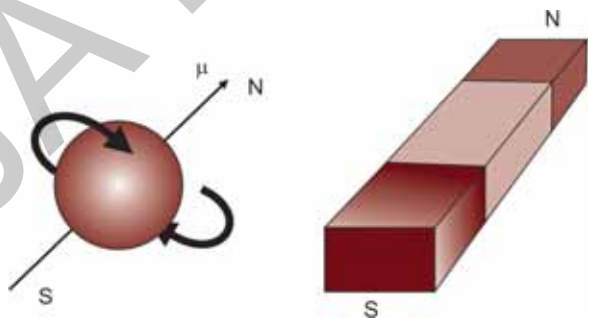
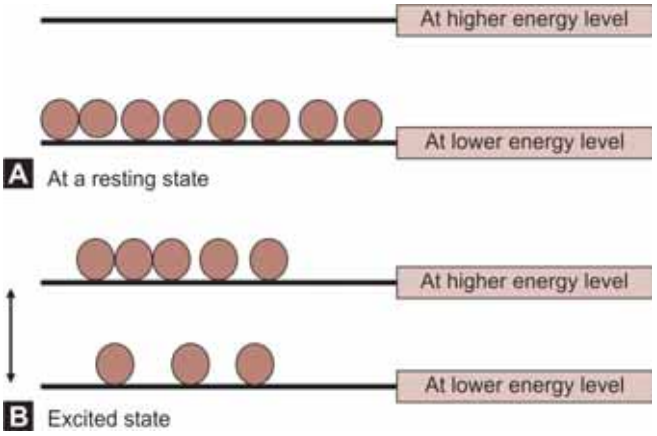
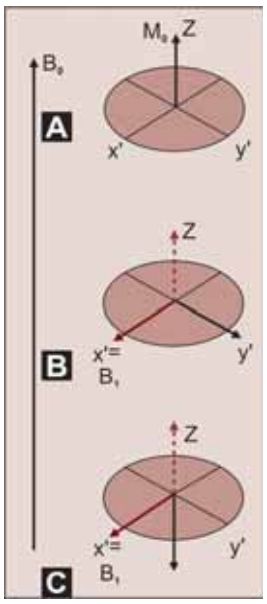


Fig. 11: Every spinning particle possesses a magnetic moment (μ) and creates a magnetic filed similar to a bar magnet



Figs 12A and B: (A) Showing protons outside a magnetic field. (B) showing excited protons in a magnetic field moving from a lower energy level to a higher energy level with two distinct energy levels. The population difference is directly proportional to the magnetic field strength



Figs 13A to C: (A) Alignment of the protons along the direction of the external magnetic field (B_0) in the z-axis. (B) After applying the RF pulse of an appropriate frequency, the magnetisation (M_0)/ protons are tipped away from its equilibrium in the x-y plane. (C) If a longer pulse lasting twice as long is applied, the magnetisation is inverted

T1 and T2 Relaxation

When the RF pulse is switched off, two processes take place simultaneously

A. Recovery of the net magnetic moment in the Z axis—termed as longitudinal or T1 relaxation. T1 is the time required for the buildup of 63% of the original magnetisation along the Z axis (Figs 13A to C).
B. Loss of phase coherence in the X-Y plane or transverse plane—termed as T2 relaxation.

The nuclei while returning to the ground state dissipate their excess energy to their surroundings, which is called the lattice. This process is named as spin-lattice relaxation (Fig. 14). Smaller molecules reorient more rapidly than larger molecules. The medium-sized molecules, such as lipids, relax faster as their frequency

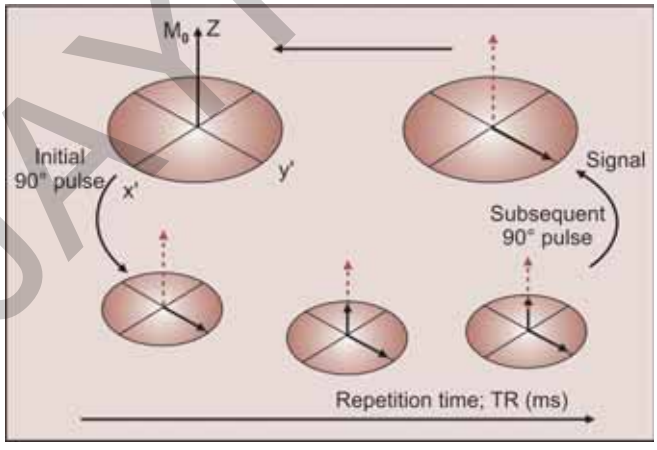


Fig. 14: Spin-lattice relaxation time

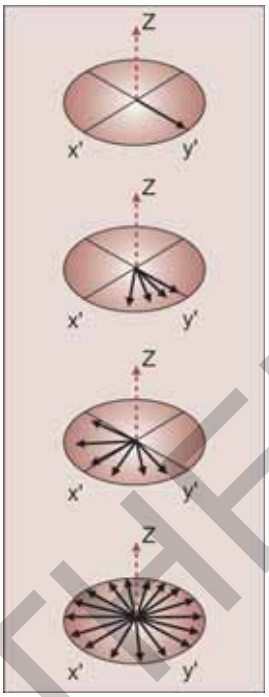


Fig. 15: Spin – spin relaxation–T2 relaxation-Loss of magnetisation in the x'-y' plane is faster than the loss of magnetisation in the z- direction due to loss of phase coherence of the microscopic components

of rotation is closer to the Larmor frequency than that associated with pure water or larger molecules such as proteins. Thus, T1 relaxation times depend on magnetic field strength because the latter affects the Larmor frequency. Thus water has a long T1.

Transverse magnetisation occurs because the magnetic field generated by the surrounding electrons exposes the precessing nuclei to different field strengths. Loss of transverse magnetisation (phase coherence) occurs as the magnetic moments get out of phase as a result of their mutual interaction. Anything that changes the magnetic field strength also changes the precessional frequency and causes a loss of phase coherence (dephasing) and shrinking of the transverse magnetisation. This is called T2 relaxation or spin-spin relaxation (Fig. 15). It denotes the loss of phase coherence caused by interactions between neighbouring magnetic moments. T2 is the time required to reduce the transverse magnetisation to 37% of its original value.

In biological tissues, the main contribution to T2 relaxation is from the relatively static magnetic field from neighbouring protons. Large molecules, which tend to reorient more slowly than small molecules, promote T2 relaxation and have shorter T2 times. Free water has a longer T2 than water associated with macromolecules. The T2 is relatively independent of the field strength.

Repetition Time

The time between two RF excitation pulses is called the repetition time (TR). The TR can be chosen from a certain minimum value, depending on the imaging technique and the MR system, to very long times.

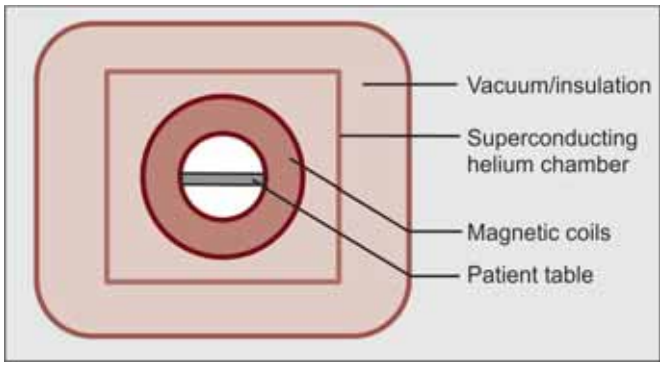


Fig. 16: Showing schematic representation of the superconducting MR systems. The bore is surrounded by the coils of the wire through which electric current is passed and cooled by liquid helium to achieve magnetisation and desired field strength

Longer values of TR allow more T1 relaxation to occur, and this property can be exploited to manipulate the contrast between tissues with different T1s or the signal-to-noise ratio in an image.

Echo Time

The time from the centre of the RF excitation pulse to the centre of the echo is the echo time (TE). The amplitude of the transverse magnetisation at the echo peak depends on TE and T2 of the tissue. As TE is prolonged, the transverse magnetisation becomes weaker. Adjusting TE influences the contrast between tissues that have different T2s.

Slice Orientation

The orientation of a slice, i.e. axial, coronal or sagittal, depends on which of the three magnetic field gradients is activated during the RF pulse. An RF pulse in the presence of the z gradient creates a transverse slice. The x and y gradients select slices in the sagittal and coronal orientations, respectively. Oblique slices are created by activating two or more gradients during an RF pulse.

Slice Position

Slices are located where the Larmor frequency matches the frequency of the RF pulse. The slice-selection gradient lowers the Larmor frequency on one side of the centre of the magnet and raises it on the other side. Slice position is controlled by changing the frequency of the RF pulse because changing the amplitude of the slice-selection gradient would inadvertently alter the thickness of the slice.

INSTRUMENTATION

The key components of an MR system are the magnet, the gradient, the RF subsystem and the computer.

The Magnet

The magnet is the main component of the MR system. There are three types of magnets in common use for

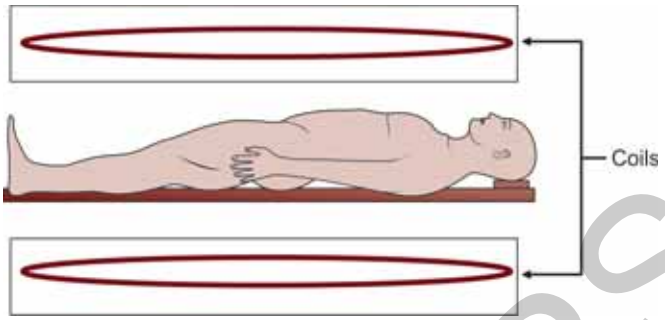


Fig. 17: Schematic diagram of a permanent MR system showing the generation of the magnetic field in a vertical direction by magnetised ceramic blocks

MRI—permanent magnets, resistive electromagnets and superconducting electromagnets. The higher the field strength the better is the signal-to-noise ratio. The strength of the magnetic field is measured in Gauss (G) or Tesla (T) units (10,000 G = 1 T). Diagnostic MR systems usually employ magnets with operating field strengths ranging from 0.02 to 3 T. Research systems operate above 3T up to 9T.

Superconducting Magnets

These are the most commonly used magnets and operate at field strength above 0.5 T. Some metals (e.g. Hg) and alloys (e.g. niobium/titanium, Nb/Ti; niobium/tin, Nb3Sn; and vanadium/gallium, V3Ga) lose their electrical resistance at very low temperatures and become superconductors. The superconductor most widely used in the construction of clinical magnets is Nb/Ti. This alloy becomes superconducting at 10° Kelvin (K) in the absence of an external magnetic field. This temperature is provided by a bath of liquid helium (4° K) (Fig. 16).

Resistive Magnets

A resistive magnet is an electromagnet in which the magnetic field is generated by the passage of electrical current through a wire. The disadvantage is their high-power consumption, limiting field strength.

Permanent Magnets

It uses a horse-shoe magnet. An advantage of these low field permanent magnet systems is that their C-shaped design is patient friendly and therefore useful in claustrophobic patients. Their field strength is limited to 0.5 T (Fig. 17).

Magnetic Field Gradients

Magnetic field gradients are activated as pulses for a short duration at timed intervals. It is a magnetic field that increases in strength along a particular direction, e.g. x, y and z gradients, according to the direction of change of the magnetic field strength. The strength of a gradient refers to the rate at which its magnetic field changes with distance.

Radiofrequency System

The excitation of the nuclei is done with a short duration RF pulse close to or at the Larmor frequency of the nuclei. The desired frequency is produced by a frequency synthesizer. The receiver detects signals in the high and very high frequency (HF and VHF) range. The magnetic resonance signals are typically a few μV in amplitude.

Transmitter and Receiver Coils

The body part to be examined is placed inside a coil. Separate coils can be used for transmitting and receiving or a single coil can be used for both excitation and detection (transceiver coil). A coil is a winding of low-resistance wire, usually made of copper. Volume coils are used for large body parts. Surface coils are used to study small regions such as the eye. The advantage of surface coils is that their signal-to-noise ratio is better as the part is close to the coil. Surface coils can receive a good signal from the tissues within the depth of half its diameter.

COMMONLY USED PULSE SEQUENCES

Spin-Echo Pulse Sequence

In a spin-echo pulse sequence two RF pulses, i.e. 90° and 180°, are applied spaced by a time interval of TE/2. After the nuclei are excited by a 90° pulse, the spins dephase in the x′-y′ plane and this is followed by a refocusing 180° pulse. The faster spins lie behind the slower ones, but at time TE/2 they make up, thus producing an echo.

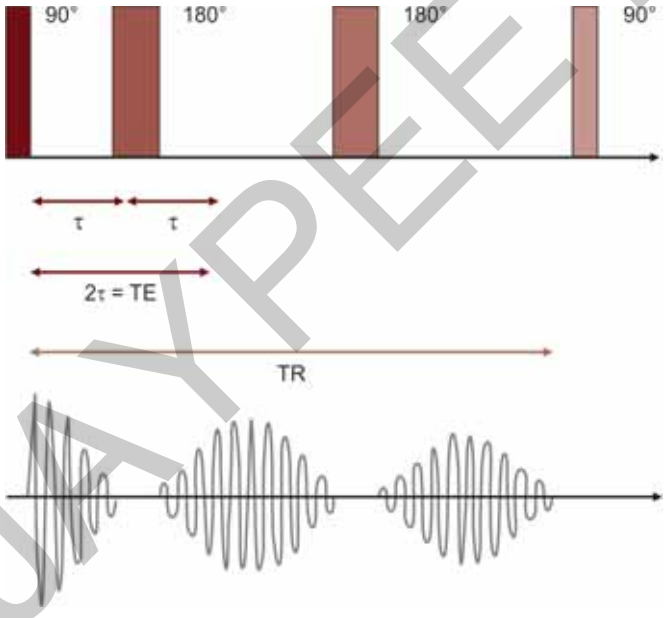


Fig. 18: Diagram of a spin-echo pulse sequence. The spin system is excited by a 90° pulse. After a time delay, one or several 180° pulses follow. This leads to the formation of an echo. The time between the 90° pulse and the peak of the echo is called echo time (TE). TR is the repetition time between two complete pulse sequences

The 180° pulse results in reversal of the phase of each spin. The position of the spins has not changed, so they will continue to rotate in the same direction. However, the 180° pulse causes the spins to return towards their starting point, rather than rotating further away from it. This 90°–180° pulse sequence is called spin-echo sequence (Fig. 18). By altering the echo delay time, and the sequence TR, the spin-echo sequence can be used to obtain T1, T2 or proton density images. The spin-echo sequence has been largely replaced by faster sequences such as fast spin echo and fast gradient recalled echo (GRE).

Gradient Echo Imaging

Gradient echo imaging is an imaging technique by which images can be acquired in much shorter times than conventional pulse sequences. The basic difference between spin-echo and gradient echo imaging is that gradient echo uses gradient reversals to get an echo, and spin echo uses 180° rephrasing pulse and gradient echo uses flip angle less than 90° (Fig. 19).

Inversion Recovery Imaging

The inversion recovery sequence uses a 180° inverting pulse, a 90° pulse and a rephrasing 180° pulse. The inversion time (TI) is determined by the TR and T1 of the tissue needed to be suppressed (Fig. 20). Commonly used inversion recovery pulse sequence are:

1. Fluid attenuated inversion recovery (FLAIR) whereby the cerebrospinal fluid (CSF) bright signal is suppressed. It is now a routinely used sequence in

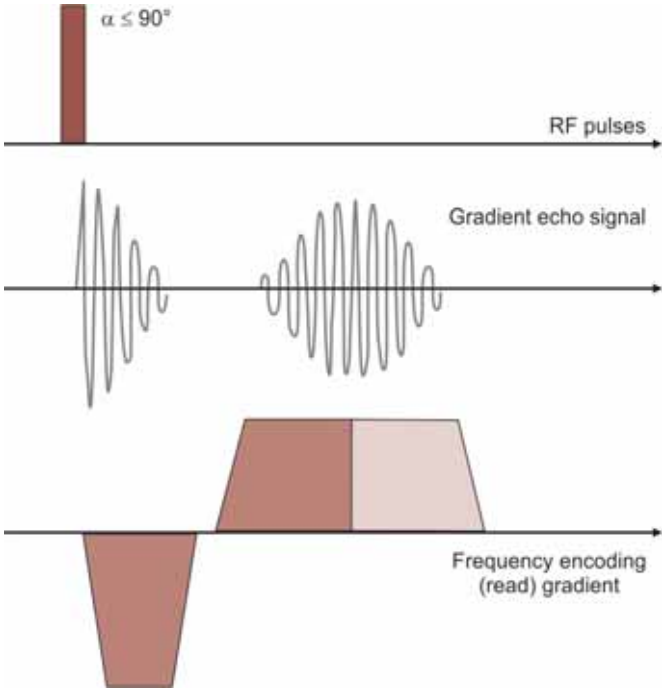


Fig. 19: Formation of a gradient echo. Instead of the 180° pulse, a gradient pulse (-G) is used followed by a second gradient pulse of opposite polarity (+G). In gradient echo sequence, the signal decay is determined by T2*, which is always less than T2

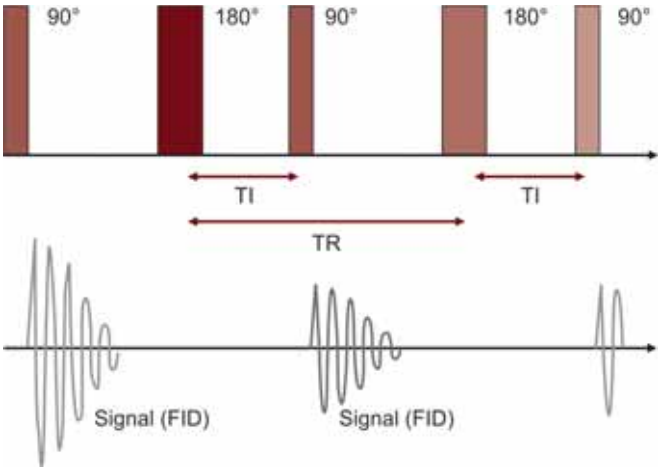


Fig. 20: Pulse sequence diagram of an inversion recovery pulse sequence. The 180° inverting pulse is followed by a 90° pulse and 180° rephasing pulse

brain imaging and especially to image periventricular plaques in multiple sclerosis.

2. Short tau inversion recovery (STIR) sequence is mainly used in imaging the optic nerves. It suppresses the orbital fat and highlights the lesions within the optic nerve, mainly in optic neuritis.

MAGNETIC RESONANCE CONTRAST

Most of the contrast agents in clinical use enhance tissue relaxation. Gadolinium is a rare earth element and toxic by itself, hence it is chelated with multi-dentate ligands for safety such as diethylenetriamine pentetate (DTPA) and tetraazacyclododecane tetraacetic acid (DOTA). It is a paramagnetic substance that shortens the T1 relaxation and hence makes the tissues with contrast appear bright.

Safety

- These contrast agents are considered safe with a rate of adverse reaction such as nausea and vomiting (1–2%) and hives (1%). Severe anaphylactoid reactions have been reported with an estimated rate of 1 in 200,000 and 1 in 400,000.
- These contrast agents can be safely used in children above 2 years.
- They should not be used in patients with compromised renal function. There have been cases reported of nephrogenic systemic fibrosis in patients with compromised renal function.
- Should not be used in pregnancy as its bioeffect on the foetus has not been established.

MAGNETIC RESONANCE ANGIOGRAPHY

Advantages of magnetic resonance angiography (MRA) versus catheter angiogram are:

- Non-invasive or minimally invasive
- Three-dimensional information can be obtained

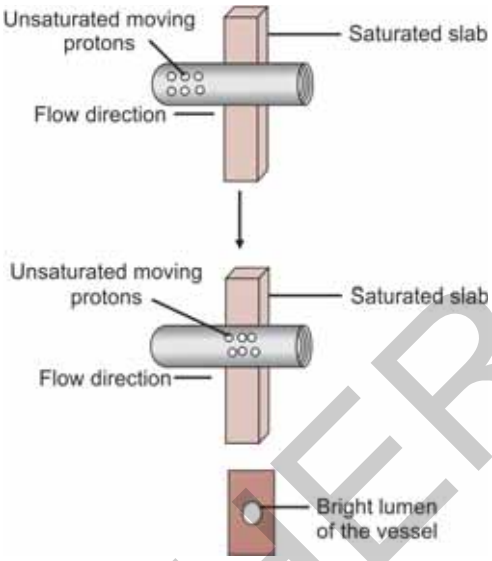


Fig. 21: Schematic representation of time-of-flight angiogram

- Can give surrounding soft tissue details

Disadvantages include:

- Flow dynamic information is lacking

Techniques of Magnetic Resonance Angiography

The commonly used techniques in clinical practice are:

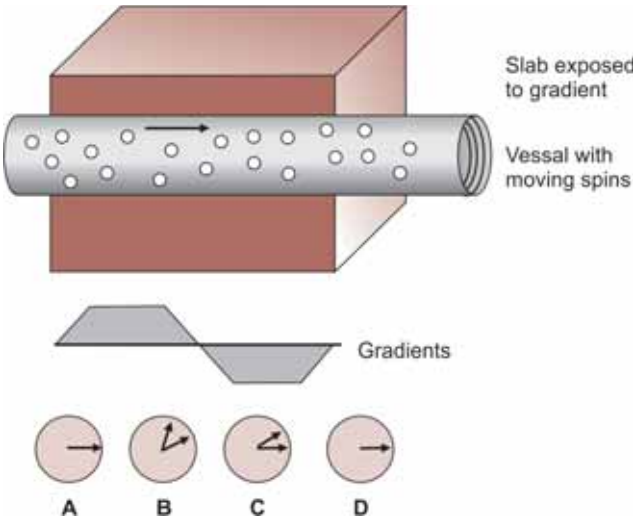
- Time-of-flight (TOF) MR angiogram
- Phase contrast (PC) MR angiogram
- Contrast-enhanced (CE) MR angiogram

Time-of-Flight Magnetic Resonance Angiogram

This is the most widely used MR angiography technique for imaging the intracranial circulation. It gives reliable vascular information without the need for intravenous contrast.

The basic principle involves suppression of the static background tissue and retaining the signal from the flowing blood. The saturation of the stationary tissue is done by using very short TR so that the stationary spins do not have enough time to regain their longitudinal magnetisation. The flowing (unsaturated) spins which enter the slice are unaffected by the slice selective RF pulse and will be fully magnetised producing a bright signal (Fig. 21). The signal produced is directly proportional to the velocity of the flowing blood. Flow saturation will occur when the spins in the imaging volume are not entirely replenished after each pulse.

The TOF angiogram can be obtained using 2-D or 3-D sequences. In a 2-D sequence, sequential thin sections are obtained whereas in 3-D a slab of tissue is excited. Each of them has their advantages and disadvantages. Two-dimensional angiograms are used to evaluate slow flowing blood, but are susceptible to turbulent flow. There is less spatial resolution. Three-dimensional angiograms have high spatial resolutions and are less susceptible to turbulent flow.



Figs 22A to D: Schematic diagram of a phase contrast MR angiogram. (A) Spins in the stationary tissue at time 0. (B) Spins dephasing after exposed to gradients. (C) Spins rephasing after switching off gradient. (D) Stationary spins rephased while moving spins are out of phase

Phase Contrast Magnetic Resonance Angiogram

Moving spins undergo a phase shift in the presence of paired opposing gradients. This phenomenon is utilised in phase contrast magnetic resonance angiogram (PC MRA) The amount of phase shift increases with increasing flow velocity. When the flowing blood (moving spins) moves along the direction of the gradient field, it precesses faster as the field increases and undergoes a phase change. Thus, the motion is phase encoded giving it both direction and magnitude (Figs 22A to D).

The amount of phase shift is directly proportional to the flow velocity, gradient strength and time interval between the gradient applications. By choosing an appropriate velocity encoding value (VENC), fast or slow flowing blood can be imaged. Phase contrast MRA can be acquired as both 2-D and 3-D sequences.

Advantages of phase contrast magnetic resonance angiogram: It gives:

- Flow quantification
- Flow direction
- Excellent background suppression
- Can be used for imaging areas of slow flow

Disadvantages of phase contrast magnetic resonance angiogram: The disadvantage is as follows:

- Long scan time.

Contrast-Enhanced Magnetic Resonance Angiography

The limitations of TOF and PC angiograms, such as flow saturation, flow-related artifacts, breathing and pulsation artifacts, made depiction of blood vessels in the body, especially the abdomen, difficult. By using intravenous contrast and rapid gradient imaging, it is now possible to obtain MRA images almost at par with conventional angiogram. The technique involves capturing of high magnetisation strength during the first pass of the vascular contrast, i.e. gadolinium, by appropriate timing using 3-D acquisition (Fig. 23).

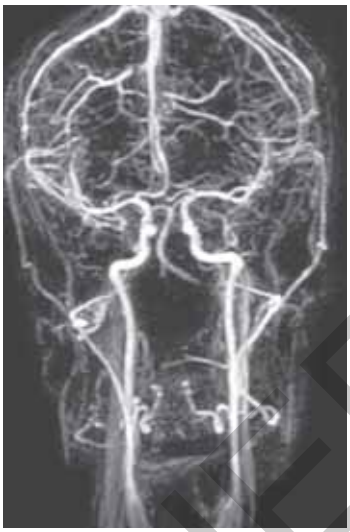


Fig. 23: Contrast- enhanced time-resolved imaging of contrast kinetics angiogram image of the brain

Advantages: The advantages are as follows:

- Insensitive to saturation effects of the RF pulse as against TOF angiogram and therefore can cover vessels over a larger FOV.
- Useful in large aneurysms where flow is complex.

NEWER ADVANCED MAGNETIC RESONANCE IMAGING TECHNIQUES

Diffusion-Weighted Imaging

It is based on the principle of Brownian motion, which is dispersion or random translation of a molecule in a liquid due to thermal agitation.

Motion of molecules in biological tissues is complex. Neuronal tissue consists of tightly and coherently packed axons surrounded by glial cells. The movement of water molecules is hindered in a direction perpendicular to the orientation of the axonal fibres. Thus motion of molecules in biological tissues is anisotropic. The cell membranes are thought to be responsible for anisotropic diffusion rather than myelin. The restricted diffusion appears as a bright signal on diffusion-weighted images (Fig. 24).

Applications

- Stroke
- Multiple sclerosis
- Tumours
- Trauma
- Abscess

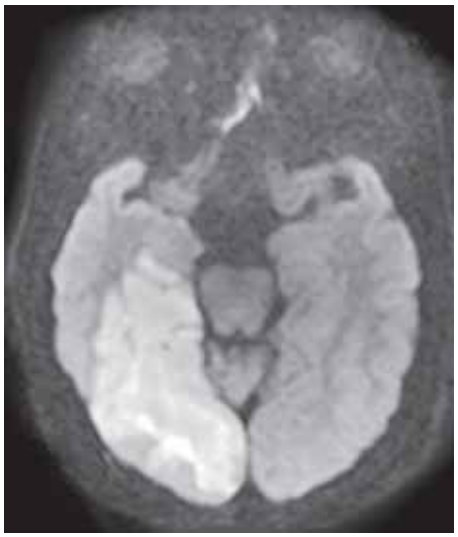


Fig. 24: Axial diffusion-weighted image showing restricted diffusion in the right occipital lobe suggestive of acute right posterior cerebral artery infarct

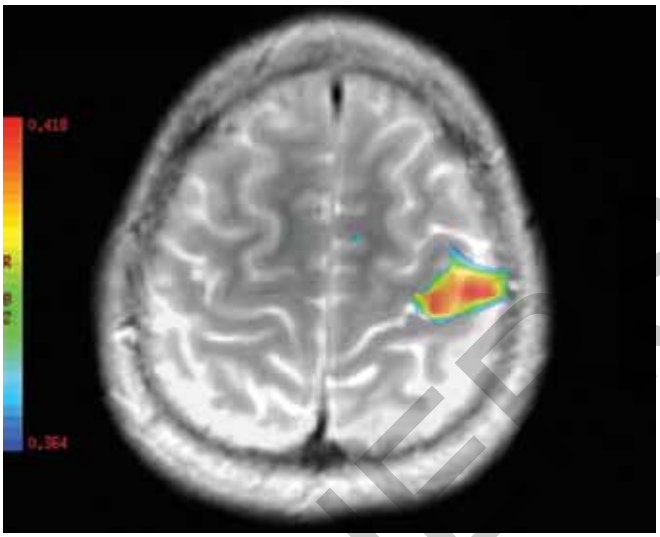


Fig. 25: MRI showing activation of the left motor cortex after right finger tapping

Lesions Bright on Diffusion Images

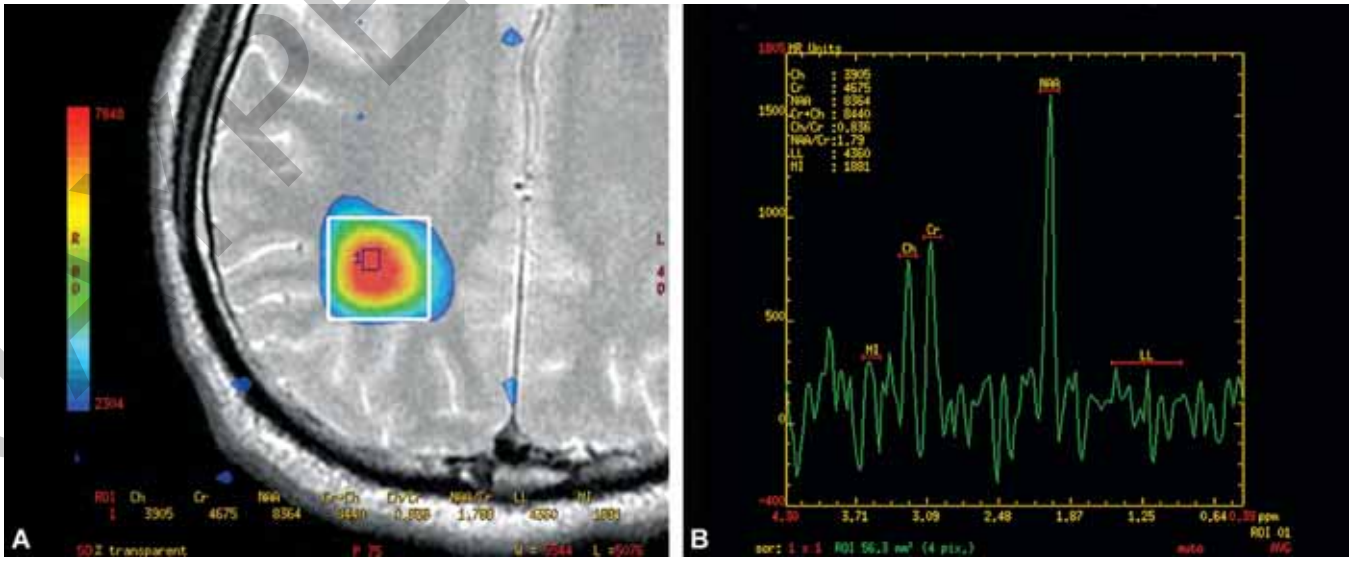
- Acute infarct
- Bacterial abscess
- Acute demyelination
- Epidermoid cyst
- Tissues with high cellularity
- Subacute haemorrhage

Functional Imaging

It is the demonstration of brain activation to a specific stimulus based on the functional anatomy of the brain, e.g. the primary visual cortex is activated using a flicker display or alternating checkerboard pattern as a visual stimulus. Once the brain is activated using a stimulus, there is change in the blood flow to the particular region due to the increased demand for oxygen and glucose. This increase in oxygen, i.e. deoxyhaemoglobin concentration causes local susceptibility effects which are used to receive the signals using appropriate pulse sequences. This is termed as blood oxygen level-dependent (BOLD) contrast imaging (Fig. 25).

Magnetic Resonance Spectroscopy

Magnetic resonance spectroscopy utilises the differences in the resonance frequency of nuclei due to their different chemical bond. This is also termed as chemical shift imaging. The frequency difference varies with the magnetic field and is directly proportional to the external magnetic field. It is expressed in parts per million (ppm). The advantages of higher field strength, while performing spectroscopy, are that it provides better signal-to-noise ratio and better separation of metabolite peaks.



Figs 26A and B: Multi-voxel MR spectroscopy TE=1044 ms (A) Showing the voxel placed in the normal parietal white matter with NAA colour map. (B) Showing normal spectrum

1H (proton) spectroscopy is used for brain imaging as it is easy to perform and gives a better signal-to-noise ratio as compared to 23Na and 31P. Of all the atomic nuclei, 1H has the strongest response and is found in all biochemicals. MR spectroscopy thus provides details of the brain chemistry (Figs 26A and B). The spectrum is read from right to left and the metabolites detected on brain spectroscopy are:

- Lipid 0.9–1.4 ppm
- Lactate 1.3 ppm
- N-acetyl aspartate (NAA) at 2 ppm
- Creatine (Cr) 3.0 ppm
- Choline (Cho) 3.2 ppm
- Myo-inositol 3.5 ppm

The TE affects the metabolites detected, thus short TE ~30 ms shows metabolites with short and long T2 relaxation times and with long TE ~ 270 ms only metabolites with long T2 relaxation times are detected, therefore the spectrum primarily consists of NAA, Cr and Cho. Another advantage of long TE ~ 144 ms is that the lactate peak at 1.3 ppm gets inverted. Rather than absolute concentrations, one should rely on the various ratios to give a clinical diagnosis.

Ratio	Normal	Abnormal
NAA/Cr	2.0	< 1.6
NAA/Cho	1.6	< 1.2
Cho/Cr	1.2	> 1.5

Indications

- Tumours
- Radiation necrosis versus recurrence
- Infections
- Neurodegenerative disorders
- Metabolic brain disorders
- Stroke

Magnetic resonance spectroscopy should be carefully interpreted and correlated with MR images to make a final diagnosis.

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Prakash Narain Tandon received his neurosurgical training at Oslo, Norway and Montreal Neurological Institute, Montreal, Canada. He established academic department of Neurosurgery in the All India Institute of Medical Sciences, New Delhi, India. His research contributions cover a wide range of basic and applied aspects of neurosciences. These included contributions on tuberculosis of the nervous system, developmental defects of the brain, head injury, spontaneous subarachnoid haemorrhage, brain tumours and neural transplantation. He has written over 220 scientific papers, 14 monographs and a number of invited contributions to national and international books, related to neuro-oncology, neuro-otology, neuronuclear medicine, epilepsy, etc. He has steered the establishment of a series of national facilities like the Neuroinformatic Centre, Neural Transplant Unit, a Brain Bank, a national NMR facility for biomedical research, etc. He is the founder President of the NBRC Society and Chairman of its Scientific Advisory Committee. He has been elected a Fellow/Member to serve on the policy-making bodies of various Scientific Academies, Research Councils and Government Departments, both nationally and internationally. He was nominated to deliver lectures under the India-ASEAN Eminent Persons lecture series 1999 and invited to the World Economic Forum in 1999 as one of the 10 distinguished scientists from around the world. He has been a recipient of a large number of awards and honours. These include BC Roy Award for developing a speciality (1980); BC Roy Award for Eminent Medical Scientist (1993); Sir CV Raman Medal (1997); Jawaharlal Nehru Birth Centenary Award (ISCA) 1999, etc. He was Hon. Surgeon to the President of India (1977–80) and Member of Science Advisory Council to the Prime Minister (1986–89). President of India decorated him with Padma Shri in 1973 and Padma Bhushan in 1989.

Ravi Ramamurthi is a reputed neurosurgeon from Chennai, and is the Head of the Department of Dr Achanta Lakshmipathi Neurosurgical Centre, Voluntary Health Services, Chennai. He did his MBBS from Chennai in 1976, and subsequently his MS in neurosurgery from Madras Medical College and then his FRCS Ed (SN) from Edinburgh, UK in 1983. In the past he was holding eminent positions in various renowned institutions nationally as well as internationally. He was appointed as Guest Doctor, Justus Liebig University, Giessen, West Germany for training in microneurosurgery. He was assigned the designation of Registrar in Neurosurgery, Newcastle General Hospital, Newcastle UponTyne, UK. He was an Executive Member of Neurological Society of India from 1996–99 and was holding the position of Honorary Treasurer, Neurological Society of India from 2006-2009. Among his many credentials, he has written a number of papers in national as well as international journals. He presented approximately 50 papers at prestigious conferences held in India and abroad. He has been continuously holding the position of Organising Secretary and faculty of Microworkshop since 1994.

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