COMPUTED TOMOGRAPY

Limitations of X-rays

1. The super-imposition of the three-dimensional information onto a single plane makes diagnosis confusing and often difficult.
2. The photographic film usually used for making radiographs has a limited dynamic range and, therefore, only objects that have large variations in X-ray absorption relative to their surroundings will cause sufficient contrast differences on the film to be distinguished by the eye. Thus, whilst details of bony structures can be clearly seen, it is difficult to discern the shape and composition of soft tissue organs accurately.
3. In such situations, growths and abnormalities within tissue only show a very small contrast difference on the film and consequently, it is extremely difficult to detect them, even after using various injected contrast media.
4. The problem becomes even more serious while carrying out studies of the brain due to its overall shielding of the soft tissue by the dense bone of the skull.

Basic Principle of CT

- In computed tomography (CT), the picture is made by viewing the patient via X-ray imaging from numerous angles, by mathematically reconstructing the detailed structures and displaying the reconstructed image on a video monitor.
- Computed tomography differs from conventional X-ray techniques in that the pictures displayed are not photographs but are reconstructed from a large number of absorption profiles taken at regular angular intervals around a slice, with each profile being made up from a parallel set of absorption values through the object.
- In computed tomography, X-rays from a finely collimated source arc made to pass through a slice of the object or patient from a variety of directions. For directions along which the path.
- length through-tissue is longer, fewer X-rays are transmitted as compared to directions where there is less tissue attenuating the X-ray beam. In addition to the length of the tissue traversed, structures in the patient such as bone may attenuate X-rays more than a similar volume of less dense soft tissue.
- In principle, computed tomography involves the determination of attenuation characteristics for each small volume of tissue in the patient slice, which constitute the transmitted radiation intensity recorded from various irradiation directions. It is these calculated tissue attenuation characteristicsthat actually compose the CT image.

For a monochromatic X-Ray beam, the tissue attenuation characteristics can be described by,

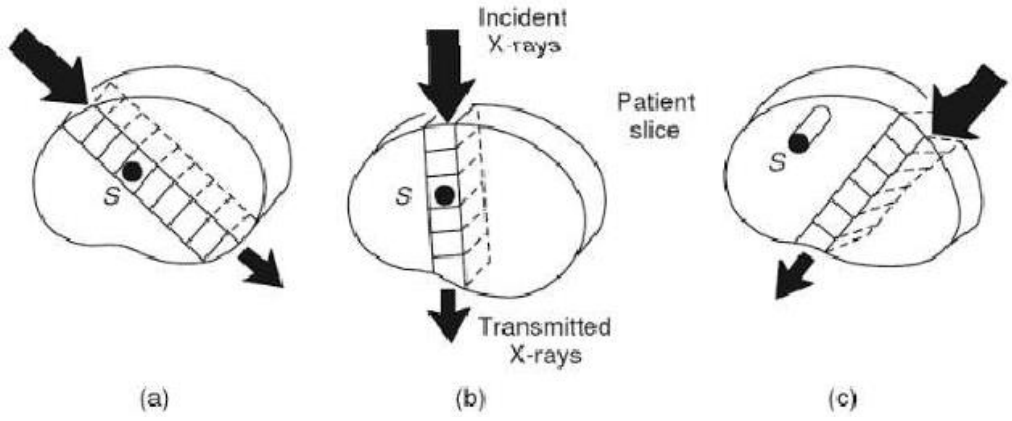
$$I_t = I_o e^{j\mu x}$$

Where,

I_o = Incident radiation intensity
I_t = Transmitted intensity
X= Thickness of tissue
μ = Characteristic attenuation coefficient of tissue

If a slice of heterogeneous tissue is irradiated given below, and we divide the slice into volume elements or voxels with each voxel having its own attenuation coefficient, it is obvious that the sum of the voxel attenuation coefficients for each X-ray beam direction can be determined from the experimentally measured beam intensities for a given voxel width. However, each individual voxel attenuation coefficient remains unknown.

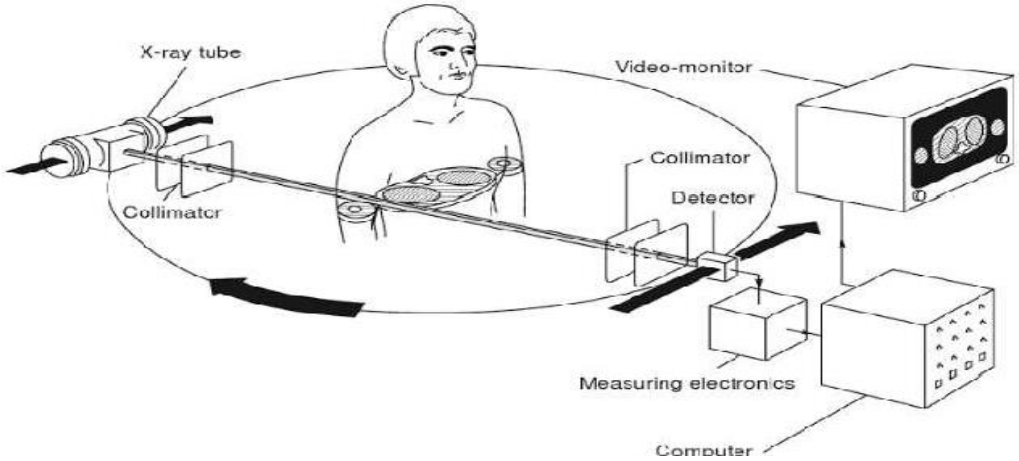
Computed tomography uses the knowledge of the attenuation coefficient sums derived from X-ray intensity measurements made at all the various irradiation directions to calculate the attenuation coefficients of each individual voxel to form the CT image



X-rays incident on patient from different directions. They are attenuated by different amounts, as indicated by the different transmitted X-ray intensities

Block Diagram of the CT System

A Block diagram of the system. The X-ray source and detectors are mounted opposite each other in a rigid gantry with the patient lying in between, and by moving one or both of these around and acrossthe relevant sections, which is how the measurements are made.



- The X-ray tube and the detector are rigidly coupled to each other. The system executes translational and rotational movement and trans radiates the patient from various angular projections. With the aid of collimators, pencil thin beam of X-ray is produced.
- A detector converts the X-radiation into an electrical signal. Measuring electronics then amplify the electrical signals and convert them into digital values. A computer then processes these values and computes them into a matrix-line density distribution pattern which is reproduced on a video monitor as a pattern of grey shade.
- In one system which employs 18 traverses in the 20s scanning cycle, 324,000 (18 x 30 x 600) X-ray transmission readings arc taken and stored by the computer. These arc obtained by integrating the outputs of the 30 detectors with approximately 600 position pulses.
- The position pulses are derived from a glass graticule that lies between a light emitting diode and photo-diode assembly that moves with the detectors. The detectors arc usually sodium- iodide crystals, which are thallium-doped to prevent an after-glow. The detectors absorb the X- ray photons and emit the energy as visible light. This is converted to electrons by a photo- multiplier tube and then amplified. Analog outputs from these tubes go through signal conditioning circuitry that amplifies, clips and shapes the signals.
- A relatively simple analog-to-digital converter then prepares the signals for the computer. Simultaneously, a separate reference detector continuously measures the intensity of the primary X-ray beam .The set of readings thus produced enables the computer to compensate for fluctuations

of X-ray intensity. Also, the reference readings taken at the end of each traverse are used to continually calibrate the detection system and the necessary correction is carriedout.

- After the initial pre-processing, the final image is put onto the system disc. This allows fordirect viewing on the operator's console. The picture is reconstructed in either a 320 x 320 matrix of 0.73 mm squares giving higher spatial resolution or in a 160x 160 matrix of 1.5 mm, squares which results in higher precision, lower noise image and better discrimination between tissues of similar density.
- Each picture clement that makes up the image matrix has a CT number, say between -1000 and + 1000, and therefore, takes up one computer word. A complete picture occupies approximately 100 K words, and up to eight such pictures can be stored on the system disc. There is a precise linear relationship between the CT numbers and the actual X-ray absorption values, and the scale is defined by air at -1000 and by water at 0.

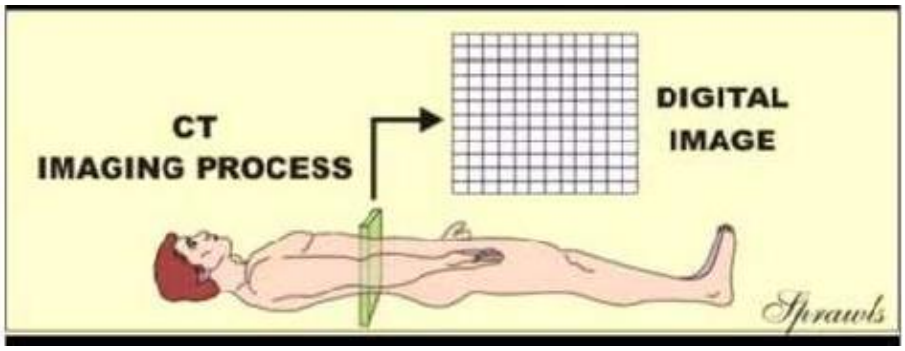
Image Reconstruction

The formation of a CT image is a distinct three phase process.

1. The scanning phase produces data, but not an image.
2. The reconstruction phase processes the acquired data and forms a digital image.
3. Digital-to analog conversion phase: The visible and displayed analog image (shades of gray) is produced by the digital-to analog conversion phase.

1. The scanning phase

- During the scanning phase a fan-shaped x-ray beam is scanned around the body. The amount of x-radiation that penetrates the body along each individual ray (pathway) through the body is measured by the detectors that intercept the x-ray beam after it passes through the body.
- The projection of the fan-shaped x-ray beam from one specific x-ray tube focal spot position produces one view. Manyviews projected from around the patient's body are required in order to acquire the necessary data to reconstruct an image.Each view produces one "profile" or line of data as shown here.
- Thecomplete scan produces a complete data set that contains sufficient information for the reconstruction of an image. In principle, one scan produces data for one slice image.



2. Image Reconstruction Phase

Image reconstruction is the phase in which the scan data set is processed to produce an image. The image is digital and consist of a matrix of pixels. Filtered back projection is the reconstruction method used in CT. "Filtered" refers to the use of the digital image processing algorithms that are used to improve image quality or change certain image quality characteristics, such as detail and noise. "Back projection" is the actual process used to produce or "reconstruct" the image

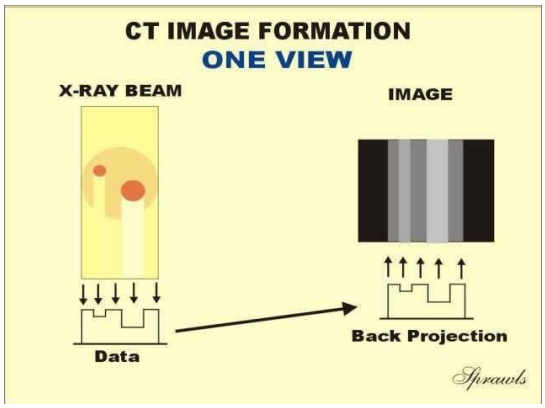
Back projection Principle

We start with one scan view through a body section (like a head) that contains two objects. As we know, the data produced is not a complete image, but a profile of the x-ray attenuation by the objects.

Let's now take this profile and attempt to draw an image by "back projecting" the profile onto our image surface.

We have now rotated the x-ray beam around the body by 900 and obtained another view. If we now back project this profile onto our image area we see the beginnings of an image showing the two object. Two views does not give us a high-quality image. Several hundred views are used to produce clinical CT

images. A part of the reconstruction process is the calculation of CT number values foreach image pixel.



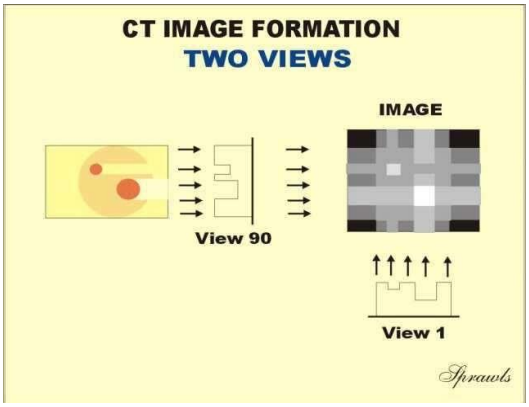
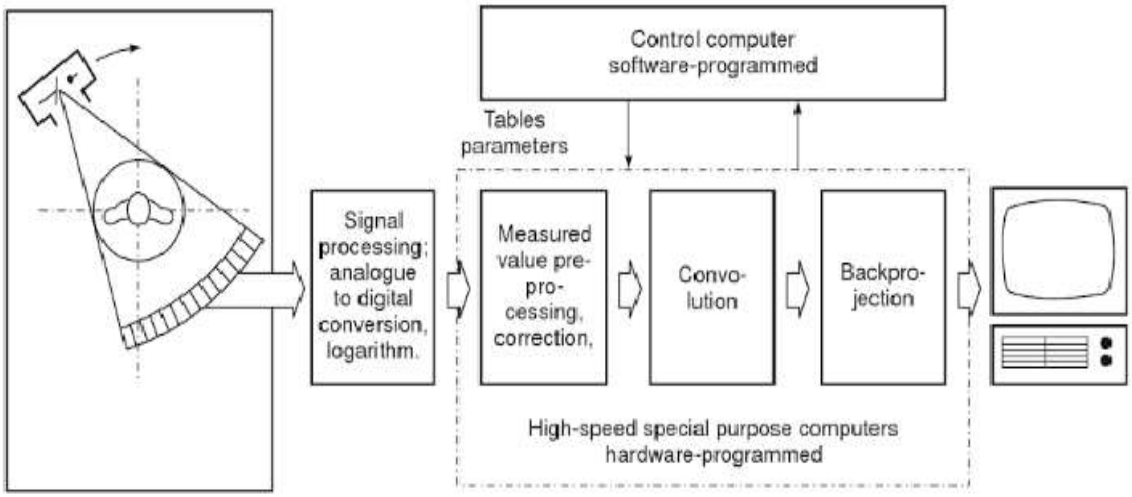


Image Reconstruction Computer, used in CT scanners.

This method enables pictures to be reconstructed within a few seconds. Figure shows a block diagram image reconstruction computer, used in CT scanners.

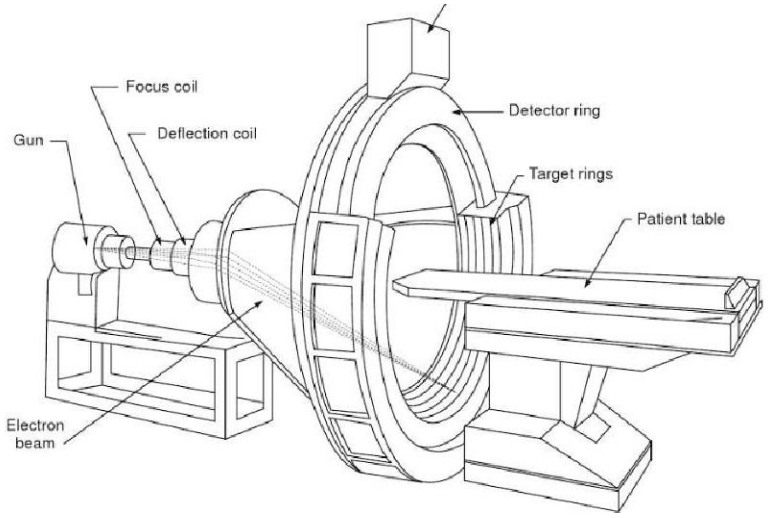


➤ Fig. 20.13 *Block diagram of the image computer. The synchronous reconstruction of the image permits the representation of the tomogram on the video monitor immediately upon completion of the scan (Courtesy: Siemen, Germany)*

Ultrafast Electron Beam CT Scanner

In this electron beam sweeps back and forth through a magnetic field. The impact of electron beam on a semi-circular tungsten array underneath the patient generates X-rays and the X ray detectors are mounted on a semi-circular array above the patient.

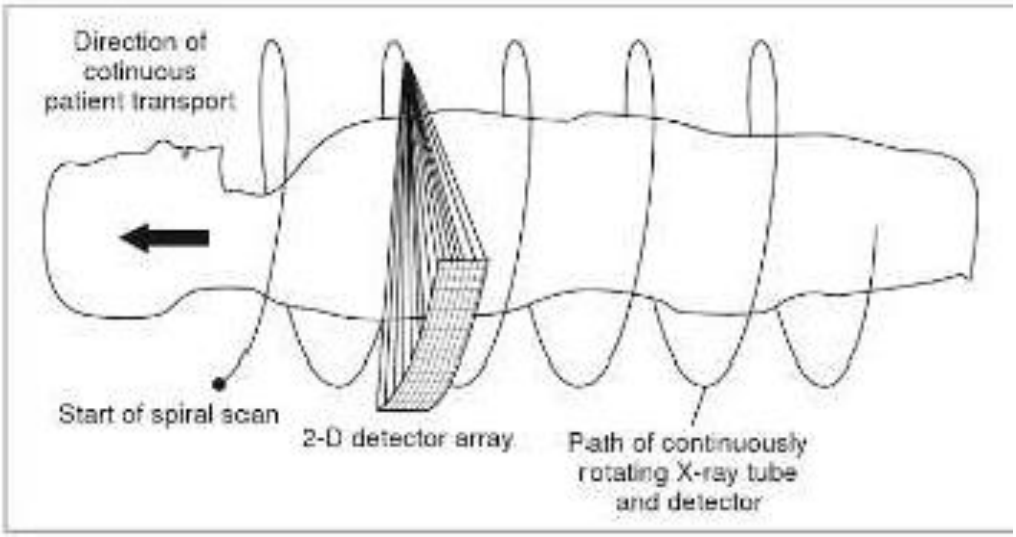
Light weight, Takes only 50ms with electron beam tomography.



- The detector array consists of two continuous ranges of 216° with 432 channels each. Luminascent crystals coupled to silicon photo-diodes are used.
- The scanning electron beam emitted by an electron gun is accelerated by 130-140 kV, electromagnetically focused and deflected over a target in a typical time of 50-100 ms.
- It was originally designed for cardiac examinations. The unit was equipped for this purpose with four anode rings and two detector rings which enabled eight contiguous slices, an area of approximately 8x8 mm. to be scanned without movement of the patient.

Spiral /Helical Scanning.

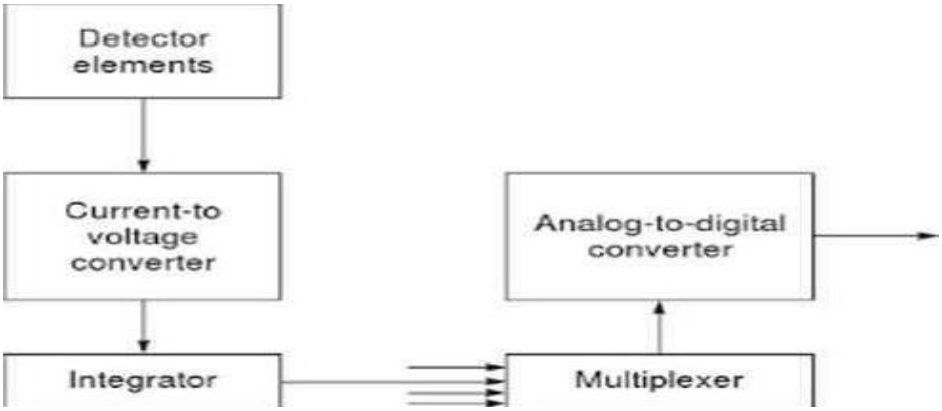
This is a scanning technique in which the X-ray tube rotates continuously around the patient while the patient is continuously translated through the fan beam. The focal spot therefore, traces a helix around the patient. The projection data thus obtained allow for the reconstruction of multiple contiguous images. This operation is often referred to as helix, spiral, volume, or three-dimensional CT scanning. This technique has been developed for acquiring images with faster scan times and to obtain fast multiple scans for three-dimensional imaging to obtain and evaluate the Volume at different locations.



The spiral scanning technique, which causes the focal spot to follow a spiral path around the patient. Multiple images are acquired while the patient is moved through the gantry in a smooth continuous motion rather than stopping for each image. The projection data for multiple images covering a volumeof the patient can be acquired in a single breath hold at rates of approximately one slice per second.

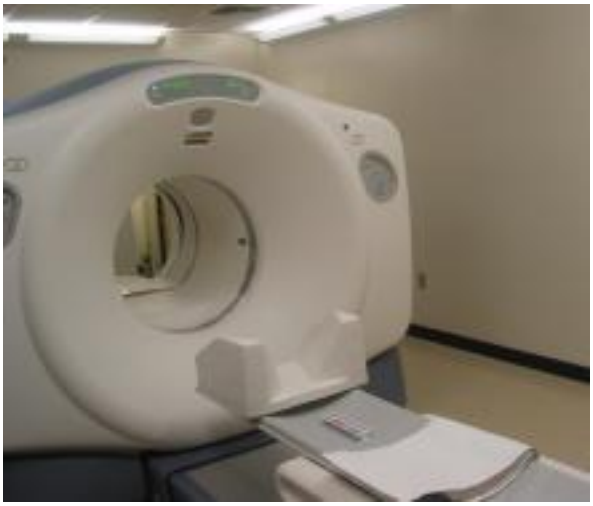
Processing System

A typical data acquisition system, it consists of precision pre-amplifiers, current to voltage convertor, analog integrators, multiplexers and analog-to-digital convertors. Data transfer rates of the order of 10 Mbytes/s are required in some scanners. This can be accomplished with a direct connection for systems having a fixed detector array. The third generation slip ring systems make use of optical transmitters onthe rotating gantry to send data to fixed optical receivers.



Magnetic Resonance Imaging (MRI)

Magnetic resonance imaging (MRI) is a spectroscopic imaging technique used in medical settings to produce images of the inside of the human body. MRI is based on the principles of nuclear magnetic resonance (NMR), which is a spectroscopic technique used to obtain microscopic chemical and physical data about molecules.



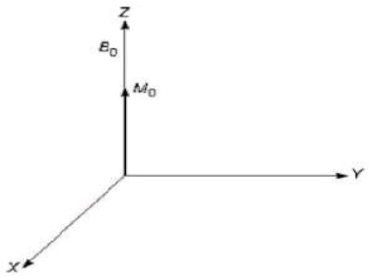
Comparison of NMR system and XRAY and CT

1. Similar to the X-ray computerized tomography (CT), MRI uses magnetic fields and radio frequency signals to obtain anatomical information about the human body as cross-sectional images in any desired direction and can easily discriminate between healthy and diseased tissue.
2. MRI images are essentially a map of the distribution density of hydrogen nuclei and parameters reflecting their motion, in cellular water and lipids.
3. The total avoidance of ionizing radiation, its lack of known hazards and the penetration of bone and air without attenuation make it a particularly attractive non-invasive imaging technique.
4. CT provides details about the bone and tissue structure of an organ whereas NMR highlights the liquid-like areas on those organs and can also be used to detect flowing liquids, like blood.
5. A conventional X-ray scanner can produce an image only at right angles to the axis of the body, whereas

the NMR scanner can produce any desired cross-section, which offers a distinct advantage to and is a big boon for the radiologist.

Basic Principle

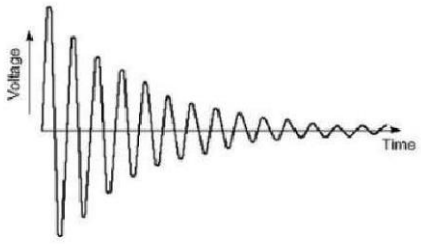
MR1 systems provide highly detailed images of tissue in the body. The systems detect and process the signals generated when hydrogen atoms, which are abundant in tissue, are placed in a strong magneticfield and excited by a resonant magnetic excitation pulse. All materials contains nucleus that have a combination of protons and neutrons. It possesses a spin and the amount of spin give rise to a magnetic moment. The magnetic moment has a magnitude and direction. In tissues Magnetic moments of nuclei making up the tissue are randomly aligned and net magnetization=0. Random alignment of magnetic moments of the nuclei making up the tissue, resulting in a zero net magnetization. When a material is placed in a magnetic field B0, some of the randomly oriented nuclei experience an external magnetic torque which tends to align the individual parallel or anti-parallel magnetic moments to the direction ofan applied magnetic field. This gives a magnetic moment that accounts for the nuclear magnetic resonance signal on which the imaging is based. This moment is in the direction of applied magneticfield Bo. With the magnetic moments being randomly oriented with respect to one another, the components in the X-Y plane cancel one another out while the Z components along the direction of the applied magnetic field add up to produce this magnetic moment M0 shown in Figure given below.



The application of external magnetic field causes the nuclear magnetic moments to align themselves, producing a net moment in the direction of the field B0

NMR Resultant Signal Pick up by the Instrument

- When a nucleus with a magnetic moment is placed in an externally applied magnetic field, the energy of the nucleus is split into lower (moment parallel with the field) and higher (anti-parallel) energy levels. The energy difference is such that a proton with specific frequency (energy) is necessary to excite a nucleus from the lower to die higher state.
- The excitation energy E obtained by the application of external RF signal, and is given by the Planck's equation E = hωo Where h is Planck's constant. This energy is usually supplied by an RF magnetic field. ωo = Frequency of appied RF.
- The excited proton tends to return or relax to its low-energy state with spontaneous decay and re-emissions of energy at a later time Y in the form of radio wave photons. This decay is exponential in nature and produces a "**free induction decay" (FID**)



Free induction decay (FID) signal obtained in NMR experiments

- To summarize, if in a static field, RF waves of the right frequency are passed through the sample of interest (or tissue), some of the parallel protons will absorb energy and be stimulated or excited to a higher energy in the anti-parallel direction. Sometime later, the RF frequency absorbed will be emitted

as electromagnetic energy of the same frequency as the RF source. The amount of energy required to flip protons from the parallel to the anti-parallel orientation is directly related to the

- Magnetic field strength; stronger fields require more energy or higher frequency radiation. This is picked up by the instrument and then processed.

BASIC NMR COMPONENTS

The basic components of an NMR imaging system are shown in Fig. These are:
1. **Magnet:** Provides a strong uniform, steady, magnet field B0.
2. **RF transmitter**, which delivers radio-frequency magnetic field to the sample.
3. **Gradient system**, which produces time-varying magnetic fields of controlled spatial non-uniformity;
4. **Detection System**, which yields the output signal; and
5. **Imager system**, including the computer, which reconstructs and displays the images.

1. Imager System

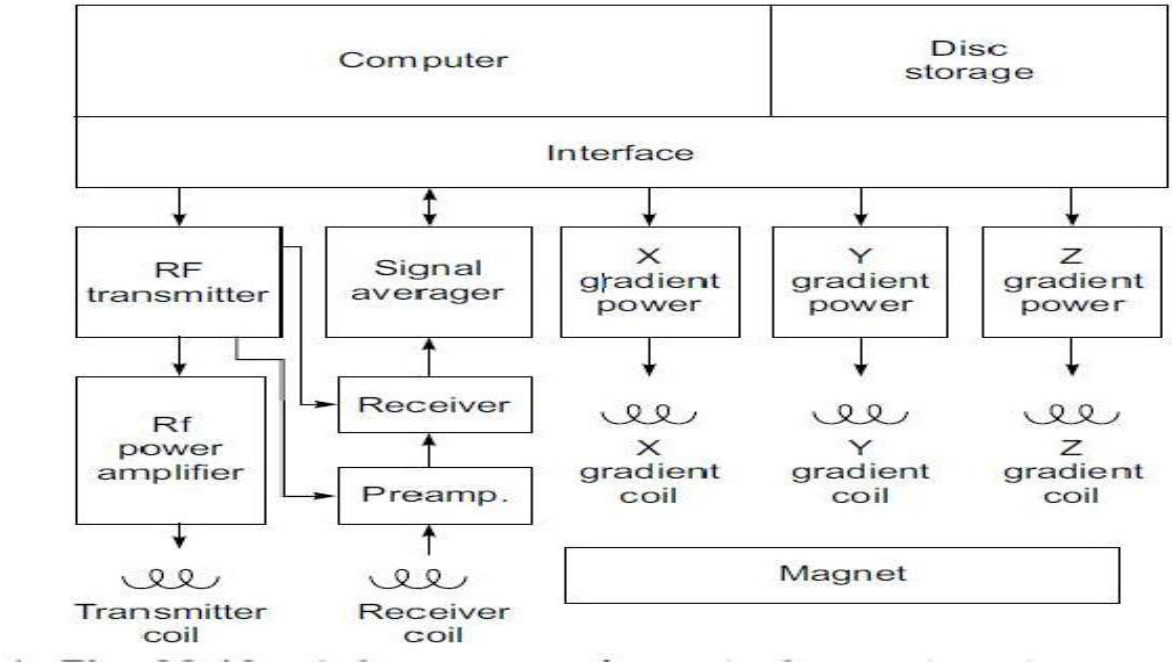
- The Imaging sequencing in the system is provided by a computer. Functions such as gates and envelopes for the NMR pulses, blanking for the pre-amplifier and RF power amplifierand voltage waveforms for the gradient magnetic fields are all under software control.

- The computer also performs the various data processing tasks including the Fourier transformation, image reconstruction, data filtering, image display and storage. Therefore, the computer must have sufficient memory and speed to handle large image arrays and data processing, in addition to interfacing facilities.

2. The Magnet:

- In magnetic resonance tomography, the base field must be extremely uniform in space and constant in time as its purpose is to align the nuclear magnets parallel to each other in the volume to be examined.
- Also, the signal-to-noise ratio increases approximately linearly with the magnetic field strength of the basic field, therefore, it must be as large as possible.
- Four factors characterize the performance of the magnets used in MR systems; viz., field strength, temporal stability, homogeneity and bore size.

- The gross non-homogeneities result in image distortion while the bore diameter limits the size of the dimension of the specimen that can be imaged. Such a magnetic field can be produced by means of four different ways, viz., permanent magnets, electromagnets, resistive magnets and super-conducting magnets.



- **Permanent Magnet:** In case of the permanent magnet, the patient is placed in the gap between a pair of permanently magnetized pole faces. Permanent magnet materials normally used in MRI scanners include high carbon iron alloys such as alnico or neodymium iron.. Although permanent magnets have the advantages of producing a relatively small fringing field and do not requirepower supplies, they tend to be very heavy (up to 100 tons) and produce relatively low fields ofthe order of 0.3 T or less.
- **Electromagnets:** Make use of soft magnetic materials such as pole faces which become

magnetized only when electric current is passed through the coils wound around them. Electromagnets obviously require external electrical power supply.

- **Resistive magnets**: make use of large current-carrying coils of aluminium strips or copper tubes. In these magnets, the electric l power requirement increases proportionately to the square of the field strength which becomes prohibitively high as the field strength increases. Moreover, the total power in the coils is converted into heat which must be dissipated by liquid cooling.
- **Superconductive magnets.** Most of the modem NMR machines utilize superconductive magnets. These magnets utilize the property of certain materials, which lose their electrical resistance fully below a specific temperature. The commonly used superconducting material is Nb Ti (Niobium Titanium) alloy for which the transition temperature lies at 9 K (-264°C). Inorder to prevent superconductivity from being destroyed by an external magnetic field or the current passing through the conductors, these conductors must be cooled down to temperatures significantly below this point, at least to half of the transition temperature. Therefore, superconductive magnet coils are cooled with liquid helium which boils at a temperature of 4.2K (-269°C).

RF Transmitter System

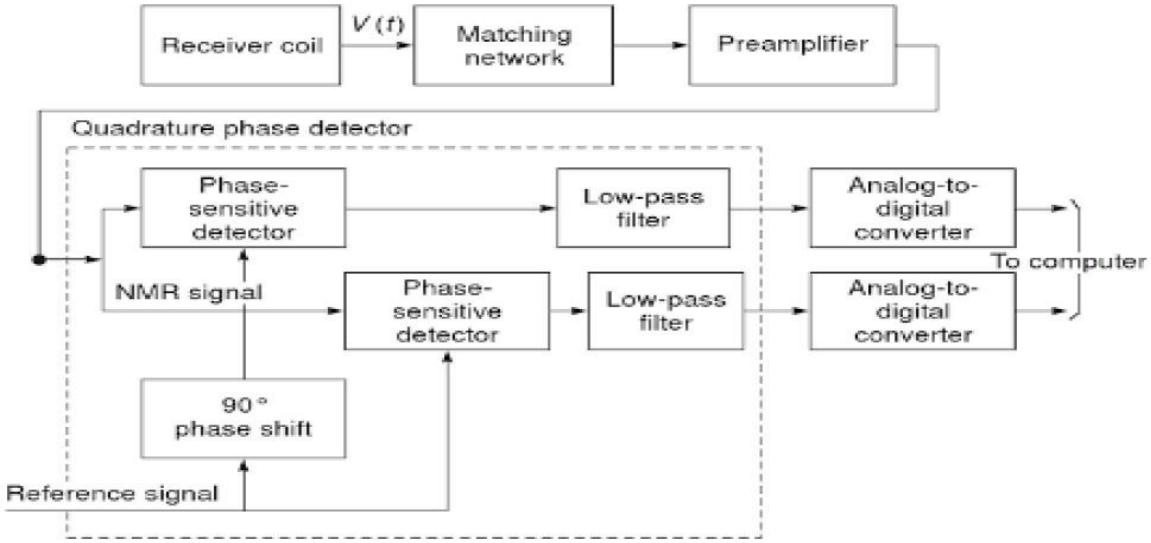
- The system consists of an RF transmitter, RF power amplifier and RF transmitting coils. 1. RF Transmitter System
- In order to activate the nuclei so that they emit a useful signal, energy must be transmitted into the sample. This is what the transmitter does.
- The RF transmitter consists of an RF crystal oscillator at the Larmor frequency. The RFvoltage is gated with the pulse envelopes from the computer interface to generate RF pulsesthat excite the resonance.

RF Power Amplifier

- These pulses are amplified to levels varying from 100W to several kW and are fed to the transmitter coil.

RF Transmitting Coils

- The coil generates RF field perpendicular to the direction of main magnetic field.
- Coils are tuned to the NMR frequency and are usually isolated from the remaining system



Block diagram of the NMR detection system

using RF shielding cage.Detection System

The function of detection system is to detect the nuclear magnetization and generate an output signal for processing by the computer.

The receiver coil usually surrounds the sample and acts as an antenna to pick up the fluctuating nuclear magnetization of the sample and converts it to a fluctuating output voltage V(i).

$$V(t) = -\frac{d}{dt} \cdot M(t,x) \cdot B_c(x)\, d_x$$

NMR signal is given by

- Where M(t, x) is the total magnetization in a volume and Bc(x) the sensitivity of the receiver coil at different points in space. Bc(x) describes the ratio of the magnetic field produced by the receiver coil to the current in the coil.
- The receiver coil design and placement is such that Bc(x) has the largest possible transverse component. The longitudinal component of Bc(x) contributes little to the output voltage and can he ignored.
- The RF signals constitute the variable measured in magnetic resonance tomography. These are extremely weak signals having amplitude in the nV (nano-Volt) range thus requiring specially designed RF antennas. The sensitivity of an MR scanner therefore depends on the quality of its RF receiving antenna. For a given sample magnetization, static magnetic field strengths and sample volume, the signal-to-noise-ratio (SN R)of the RF signal at the receiver depends in the following manner upon the RF-receiving antenna.

$$SNR \sim K(Q/V_c)$$

Where K is a numerical constant, specific to the coil geometry

Q is the coil magnetization factor, and

V_c is the coil volume.

- This implies that the SNR of an MR scan can be improved by maximizing magnetization to coil volume.
- Some of the commonly available coils are:
- Body Coils: Constructed on cylindrical coils forms with diameter ranging from 50 to 60 cm entirely surround the patient's body.
- Head Coils: Designed only for head imaging, with typical diameter of 28 cm.

Surface coils:

- Orbit/ear coil: flat, planar ring-shaped coil with 10 cm diameter;Neck coil: flexible, rectangular shaped surface coil (10 cm x 20 cm) capable of adaptation to the individual patient anatomy; and Spine coil: cylindrical or ring-shaped coil with 15 cm diameter.

Organ-enclosing coils:

Helmholtz-type coil: a pair of flat ring coils each having 15 cm diameter with distance between thetwo coils variable from 12 to 22 cm.

Matching Network

Following the receiver coil is a *matching network which* couples it to the pre-amplifier in order to maximize energy transfer into the amplifier. This network introduces a phase shifty to the phase of the signal.

Pre-amplifier:

The pre-amplifier is a low-noise amplifier which amplifies the signal and feeds it to a quadrature phase detector.

Quadrature phase detector

The detector accepts the RF NMR signal which consists of a distribution of frequencies centred around or near the transmitted frequency *w* and shifts the signal down in frequency by *w*. The detector circuit accepts the inputs, the NMR signal *V(t)* and a reference signal, and multiplies them, so that the outputis the product of the two inputs. The frequency of the reference signal is the same as that of the

irradiating RF pulse. The output of the phase-sensitive detector consists of the sum of two components, one a narrow range of frequencies centred at 2w0, and the other, a narrow range centred at zero. The low pass filter following the phase-sensitive detector removes all components except those centred at zero from the signal.

ADC

It is necessary to convert the complex (two-channel) signal to two strings of digital numbers by analog-to-digital converters. The A-D converter output is passed, in serial data form to the computerfor processing.

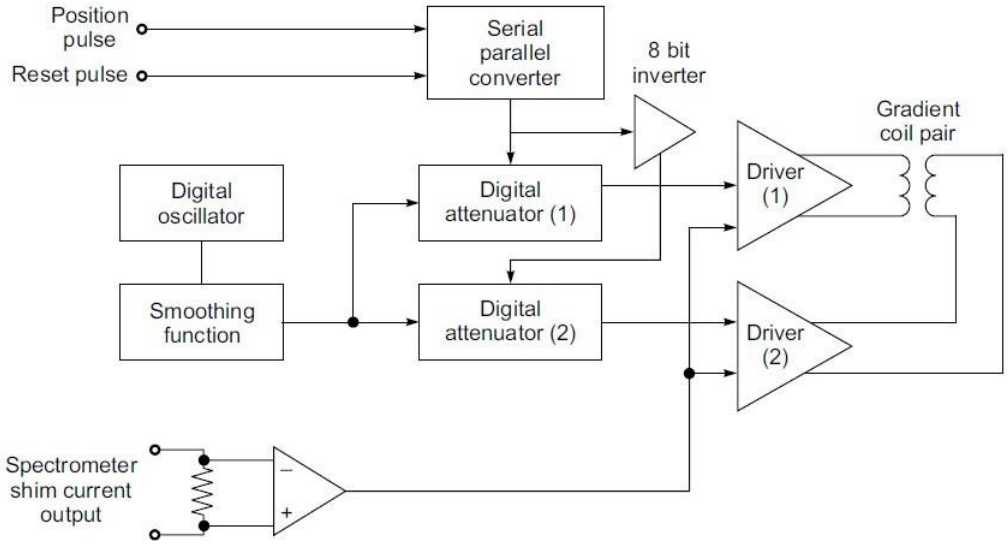
Gradient System for Spatial Coding:

Spatial distribution information can be obtained by using the fact that the resonance frequency depends on the magnetic field strength. By varying the field in a known manner through the specimen volume,it is possible to select the region of the specimen from which the information is derived on the basis of the frequency of the signal. The strength of the signal at each frequency can be interpreted as the density of the hydrogen nuclei in the plane within the object where the magnetic field corresponds to that frequency.

The imaging methods differ mainly in the nature of the gradient time dependence (static, continuously time-depended or pulsed), and in the type of NMR pulse sequence employed. Spatial information and therefore images obtained by super-imposing a linear magnetic field gradient on the uniform magnetic field applied to the object to be imaged. When this is done, the resonance frequencies of the processing nuclei will depend primarily on the positions along the direction of the magnetic gradient. This produces a one-dimensional projection of the structure of the three- dimensional object. By taking a series of these projections at different gradient orientations, a two or even three-dimensional image can be produced.

In NMR systems, for spatially resolving the signals emitted by the object, the initially homogeneous magnetic field *B0* is overlaid in all three spatial dimensions, X, Y, Z with small linear magnetic fields-gradient fields G.

These gradient fields are produced with die aid of current carrying coils and can be switched on or off as desired, both during the application of the RF energy and also in any phase of the measuringprocedure.



Serial Parallel Computer

The first sub-system includes the interface between the computer and the gradient control system. Its primary function is to allow the independent positioning of the three planes (X, *Y* and *Z).*

2. The digital oscillator

Consists of a 555-timer followed by shift registers A digital oscillator facilitates varying itis output frequency over an extremely wide range through the use of a single control.

- The 8-bit input from the interface circuit is used directly to one attenuator while the same 8-bits are inverted to control the second attenuator. The output of the attenuators is then voltage-amplified by two op amps prior to the driven circuits.
- Current control used to adjust the static field gradients be available for setting the DC levels upon which the alternating gradients are superimposed.
- An op amp serves the differential voltage drop across a dummy load and produces an output which is then DC coupled to the drivers.
- The high current drivers use a conventional design with a single op amp providing the input to a driver and a complimentary pair of power transistors to provide a sufficient current to the gradient coil.
- In typical scanners, gradient coils have an electric resistance of about l Ohm and an inductance of 1 mH. The gradient fields are required to be switched from 0 to 10 mT/ m in about 0.5 ms. the current switches from O to about 100 A in this interval. The power dissipation during the switching interval is about 20 kW. This places very strong demands on the power supply and it is often necessary to use water cooling to prevent overheating of the gradient coils.
- With well-designed coils, errors resulting from non-linear gradients will perhaps not be evident in a medical image since the image will remain clear and will not contain rigidly shaped objects or those with sharp edges for close comparison. But these gradient coils are usually designed to optimize linearity in the central region. Away from the centre, gradient linearity becomes progressively worse. Without restoration, the image will not give accurate information on the outer regions. Therefore, non-linear field gradients result in a geometrical distortion of the image reconstructed from projections.

Imager System

The imager system includes the computer for image processing, display system and control console.The timing and control of RF and gradient pulse sequences for relaxation time measurements and imaging, in addition to FT image reconstruction and display necessitate the use of a computer. The computer is the source of both the voltage waveforms of all gradient pulses and the envelopes of the RF pulses. A general-purpose mini-computer of the type used for a ('AT scanner is adequate for these purposes.

BIOLOGICAL EFFECTS OF NMR IMAGING

The three aspects of NMR imaging which could cause potential health hazard are:

(i) Heating due to the rf power.

A temperature increase produced in the head of NMR imaging would be about 0.3°C. Thisdoes not seem likely to pose a problem.

(ii) Static magnetic field:

No significant effects of the static field with die level used in NMR are known, but the possible side effects of electromagnetic fields are decrease in cognitive skills, mitotic delay in slime moulds, delayed wound healing and elevated serum triglycerides.

(iii) Electric current induction due to rapid change in magnetic field:

It is believed that oscillating magnetic field gradients may induce electric currents strong enough to cause ventricular fibrillation. However, no damage due to NMR

ADVANTAGES OF NMR IMAGING SYSTEM

1. The NMR provides substantial contrast between soft tissues that are nearly identical.

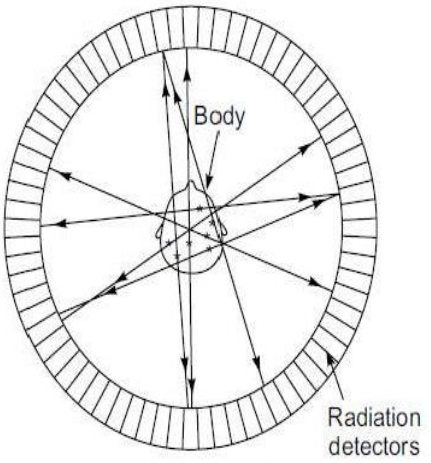
2. NMR uses no ionizing radiation and has minimal hazards for operators of the machines and for patients.
3. Unlike CT, NMR imaging requires no moving parts, gantries or sophisticated crystal detectors.
4. The system scans by superimposing electrically controlled magnetic fields consequently, scans in any pre-determined orientation are possible.
5. With the new techniques being developed, NMR permits imaging of entire three-dimensional volumes simultaneously instead of slice by slice, employed in other imaging systems.
6. In NMR both biochemical (spectroscopy) and spatial information (imaging) can be obtained without destroying the sample.

POSITRON EMISSION TOMOGRAPHY (PET) SCANNER

- Positron emission tomography is an imaging modality for obtaining in vivo cross-sectional images of positron-emitting isotopes that demonstrate biological function, physiology or pathology. Unlike anatomical imaging techniques like computed tomography (CT), X-ray, and ultrasound, PET imaging provides "functional" information about the human body. In this technique, a chemical compound with the desired biological activity is labelled with a radioactive isotope that decays by emitting a positron, or positive electrons.
- The emitted positron almost immediately combines with an electron and the two are mutually annihilated with the emission of two gamma rays. The two gamma ray photons travel in almost opposite directions, penetrate the surrounding tissue and are recorded outside the subject by a circular array of detectors
- A mathematical algorithm applied by computer rapidly reconstructs the spatial distribution of the radioactivity within the subject for a selected plane and displays the resulting image on the monitor.
- Thus, PET provides a non-invasive regional assessment of many biochemical processes that are essential to the functioning of the organ being visualized.
- The positron (β+) is emitted from a proton-rich nucleus with a variable amount of kineticenergy, the maximum amount being the endpoint energy (Eβ+), given for various isotopes

• Table.21.2 *Positron Emitters Commonly used in PET*

Isotope	$E\beta^+$, *(MeV)*	$T^{1/2}$ *(min)*
^{15}O	1.74	2.07
^{11}C	0.96	20.39
^{13}N	1.19	9.96
^{18}F	0.65	109.77
^{38}K	2.68	7.64
^{68}Ga	1.90	68.1
^{82}Rb	3.35	1.27
^{63}Zn	2.32	38.1



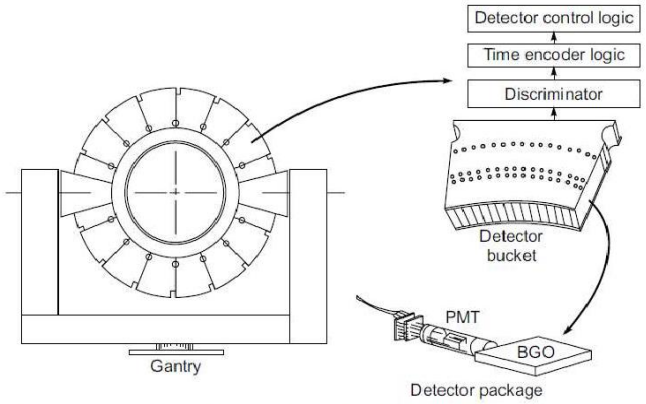
➤ Fig. 21.15 *Principle of positron emission tomography (PET) scanner*

- This energy is dissipated in the patient over a range of tissue of the order of a few millimetres. The β+ combines with a free electron (β –) and the masses are transmuted to two 511-keV g rays which are emitted at 180° ± 0.25° to one another to satisfy conservation of momentum. The variable finite range of the β+ as well as the angular variation of about 180° are

fundamental limitations to the resolution achievable with PET.

- The compounds used and quantitated are labelledwith proton-rich positron (β+) emitters that are usually cyclotron-produced. The principal isotopes are 11C, 13N, 15O, and 18F. If the compound of interest is labelled in a known position and it maintains this positron, a PETscan permits measurement of the positron concentration (μCi/mL) in a small-volume element within an organ or region of interest. This metabolic volume is typically 1 cm3.
  Two design types of positron-emission tomographs have been introduced
- One employing opposed large-area detectors which require rotation around the patient to provide the necessary degree of angular sampling another employing multiple individual crystal detectors surrounding the patient in a circular or hexagonal array.
  Conventional lead absorption:
- Collimators are not required because the coincident detection of two 511 keV photons indicatesthe line of origin along which the photons were emitted. However, in order to reduce the randomcoincidence count rate, some degree of collimation is normally employed.
- Pulse processingneeds to be much faster than with single-photon systems, to keep random coincidences tomanageable proportions. With fast-response detectors and suitably fast electronics, it maybe possible to use the difference in the time of arrival of the annihilation photons at oppositedetectors to locate the site of positron decay and improve spatial resolution.
- The gantry has a large opening and can image both the brain andtorso of adult patient.
- Axially, the tworings are separated by 36 mm. Besides containing the two BGO crystals and PMTs, the bucketalso contains amplifiers/discriminators and other front-end processing electronics.
- In order toincrease linear sampling, the entire detector assembly can wobble in a small

circular orbit. Thiswobbling procedure is used to optimize spatial resolution.



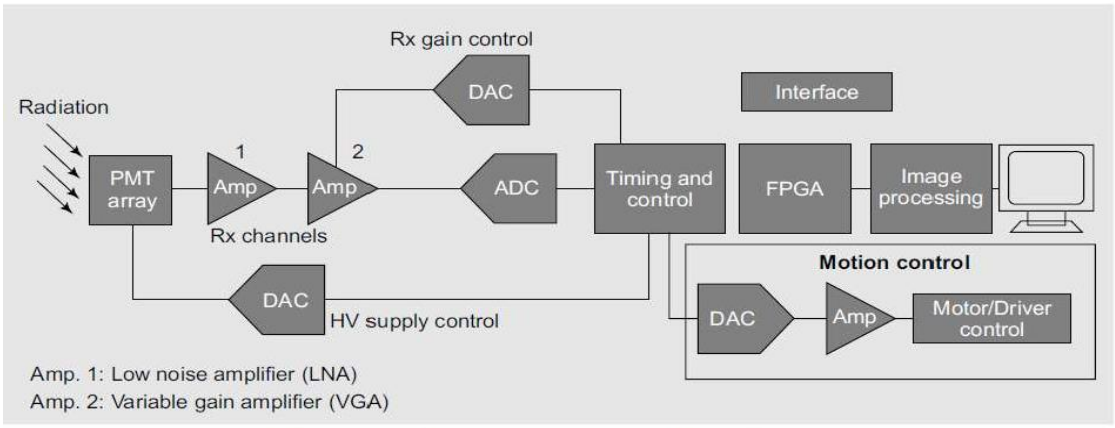
➤ Fig. 21.17 *Gantry and detector modules used in PET scanner (after Hoffman et al., 1985)*

The original PET scanners were constructed using a thallium-doped sodium iodide (NaI (TI)) detector. Its high efficiency at 511 keV, ease of fabrication, and low cost made it an obvious choice in a number of initial designs utilizing discrete crystals. Its principal disadvantage in PET work was the decreasing detection efficiency caused by the trend toward smaller crystals required for high resolution while maintaining a reasonably high total system efficiency.

A new generation of high resolution, high-efficiency PET scanners has become possible.
The use of CdZnTe (CZT) and CdTe room temperature semiconductors as detectors. CZT and CdTe can be made into large detectors for reading out resolution elements. One of the main differences between the two materials is the charge transport properties. The actual differences can favour either CZT or CdTe depending on the growth techniques used and the temperature of the devices. Most investigators have reported that the charge mobility of CdTe is somewhat better (typically on the orderof 20%) at room temperature. The devices generally have much better energy resolution than scintillator based detectors and can be made to provide very good spatial resolution.

Block diagram of a PET system.

- The PET detector is comprised of an array of thousands of scintillation crystals and hundreds of photomultiplier tubes (PMTs) arranged in a circular pattern around the patient. The current signal from each PMT output is converted to a voltage and amplified by a low-noise amplifier (LNA). The signal generated by the PMT is a pulse with a fast attack and slow decay.
- The signal strength from each PMT is determined by digitally integrating the area under this time-domain pulse. The system uses a variable-gain amplifier (VGA) after the LNA to compensate for variability in the sensitivity of the PMTs

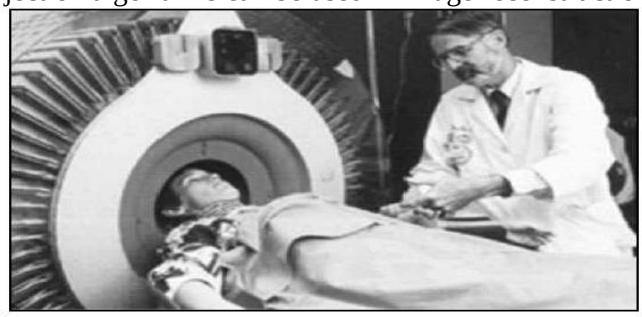


➤ Fig. 21.18 *Block diagram of a PET scanner system (Adapted from M/s Texas Instruments)*

- The combined LNA and VGA gain is approximately 40dB. The amplifiers used typically have noise of a few nV/√Hz or less, with bandwidths in the 100 kHz to 1GHz range. Current feedback amplifiers are sometimes used to provide high speed while minimizing power.
- High-density digital-to-analog converters (DACs) with 10-bit to 12-bit resolution are used to control the gain of the VGAs. The VGA's output is passed through a lowpass filter, offset compensated, and then converted to a digital signal by a 10-bit to 12-bit analog-to-digital converter (ADC) sampling at a 50 Msps to 100 Msps (mega samples/second) rate). The ADC samples are typically processed by a field programmable gate array (FPGA) discriminator which can process multiple ADC outputs.
- The signals from a number (typically four or more) of physically close PMTs are summed, and this combined signal drives the input of an ultra-high-speed comparator. A DAC generates the comparator's reference voltage to compensate for DC offsets. Extremely high accuracy is required to calculate time of flight, so a digital timestamp is generated using the comparator's output signal and an ultra-high-speed clock. In this way, timing information can be compared for multiple PMTs that are physically separated by a significant distance.
- The photon pair defines a line on which the collision took place. This is called the line ofresponse (LOR)

- Newer, higher performance PET systems are now using the time- stamps of the two photon-strike events to determine the approximate location of the collision site on the LOR. This technique improves image quality.

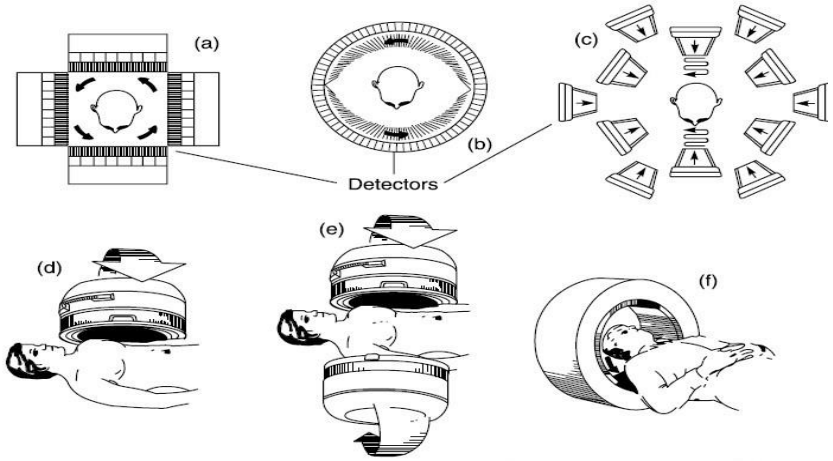
- Signal processing is needed for detector signal processing of the receive channels and for a number of control functions. Digital Signal Processors (DSPs), microcontrollers and digital-to- analog converters are used for functions such as varying input amplifier gain, controlling the PMT high-voltage power supply, and motion control for the detector ring assembly and patient entry/exit. Filtered back-projection algorithms can be used in image reconstruction.



➤ Fig. 21.19 *A typical PET system in use*

SINGLE-PHOTON-EMISSION-COMPUTED TOMOGRAPHY (SPECT)

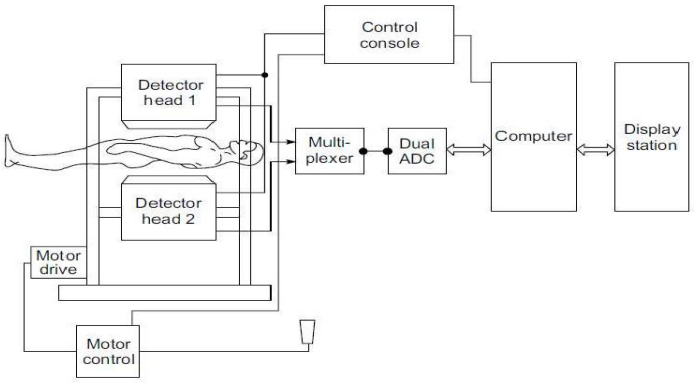
- SPECT is a nuclear imaging scan that integrates computed tomography (CT) and a radioactive tracer. The tracer is what allows doctors to see how blood flows to tissues and organs.Most stationary and some mobile gamma cameras can perform SPECT
- SPECT cameras detect only radio-nuclides that produce a cascaded emissionof single photons.
- SPECT radio-nuclides do not require an on-site cyclotron. However, the isotopes of Tc, TI,In, and Xe are not normally found in the body.
- SPECT has been used mainly in the detection of tumours and other lesions, as well asin the evaluation of myocardial function using TI-201. However, certain pharmaceuticals havebeen labelled with iodine and technetium and provide information on blood perfusion withinthe brain and the heart.
- The projection data are combined to produce transverse (also called axial or trans-axial) slices. Sagittal and coronal image slices can also be produced through mathematical manipulation of the data.
- SPECT systems with multiple camera heads are also available. In a dual-head system, two180° opposed camera heads are used, and acquisition time is reduced by half with no lossin sensitivity.
- A triple-head SPECT system further improves sensitivity. Some suppliers also offer variable-angle dual-head systems for improved positioning during cardiac, brain and whole-body imaging. Imaging times can be decreased by using another SPECT configuration—a ring of detectors completely surrounding the patient.
- Although multiple camera heads reduce acquisition time, they do not significantly shorten procedure/exam time because of factors such as patient preparation and data processing
- The sensitivity of a SPECT system is mainly determined by the total area of the detector surface that is viewing the organ of interest.



➤ Fig. 21.13 *Examples of several discrete detector and camera-based approaches for SPECT (*

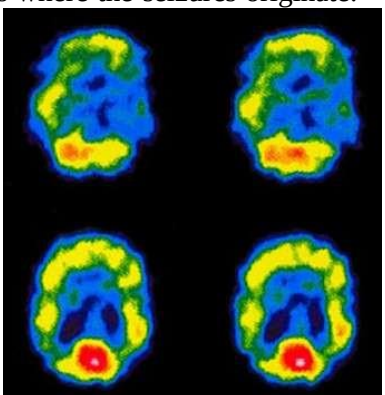
- Camera-based approaches for SPECT have the advantage of generating true three- dimensional images of the entire organ of interest. An obvious method to improve the sensitivity of these systems is to use more than a single camera
- A pallet, designed to minimize gamma ray attenuation, supports the patient between the two scintillation cameras. The camera separation is radially adjustable from 22 to 66 cm detector surface-to-surface. This adjustment range permits the collimators to be in close proximity to the patient for both body and brain scans. The data are collected with continuous gantrymotion during a 360° rotation.
- Acquisition times may be varied from 2 to 26 minutes. Angular samples are stored into 2° frames.

- The two Nal (TI) crystals, each having a useful field-of-view of 40.6 cm, are 9.5 mm thick. Each detector crystal is optically coupled to an array of 37 photo-multiplier tubes. Detector electronics include the circuitry to compensate for positioning non-linearities and regional sensitivity variations.



➤ Fig. 21.14 *Simplified diagram of SPECT system consisting of dual large field-of-view scintillation cameras mounted on a rotatable gantry*

- During acquisition, each x-y pair of gamma ray event coordinates is digitized into a 128(Perpendicular to axis-of-rotation (x)) by 64 (parallel to axis-of-rotation (y)) storage array inbuffer memory, together with a detector identifying bit and energy window identifying bit.
- Besides the primary photo-peak window, a secondary energy window is simultaneously used to record events which have undergone Compton scattering within the patient.
- Multi-slice projection set conversion and angular framing are done in real-time by the computer. The resulting projections may then be stored on disc or magnetic tape for laterimage reconstruction. The fast and common evaluation method for reconstruction of images inSPECT is by means of filtered back projection.Parallel hole collimation is used for imaging organs such as the liver, lungs and the heart.For the brain.
- Image display is accomplished on a system interfaced to a computer. A 256 ¥ 256 imageformat with 256 shades of gray with windowing and background subtraction is available.
- The image display station is directly interfaced to a film recorder. Either transparencies or rapid-process prints may be produced.
- An ECG gate is interfaced to the system. Thus, it is possible to acquire multi-gated enddiastolicand end-systolic SPECT images of the heart. Coronal and sagittal sectional images are generated from the set of contiguous transverse slices using a data re-organization algorithm.
- A SPECT scan is primarily used to view how blood flows through arteries and veins in the brain. Tests have shown that it might be more sensitive to brain injury than either MRI or CT scanning because it can detect reduced blood flow to injured sites.
- SPECT scanning is also useful for presurgical evaluation of medically uncontrolled seizures (Fig. 1). The test can be performed between seizures (interictal) or during a seizure (ictal) to determine blood flow to areas where the seizures originate.



The temporal lobe on the left side of the brain shows less blood flow than the right

- This type of scanning is also useful in diagnosing stress fractures in the spine (spondylolysis),blood deprived (ischemic) areas of brain following a stroke, and tumors.