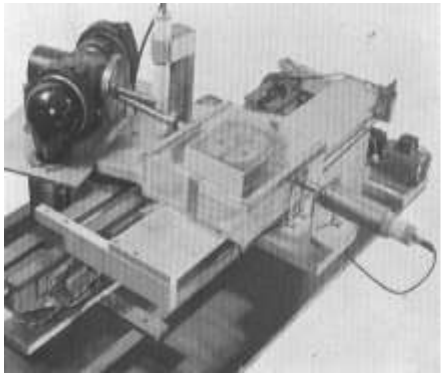
Unit II - Computed Tomography

The word "tomography" is derived from the Greek :
tomos (slice/section/part) and
graphein (to write).

Tomography:

Definition: imaging of an object by analyzing its slices.

- Mathematical principles of CT were first developed in **1917** by Johann Radon.
- **In 1924** - mathematical theory of tomographic image reconstructions (Johann Radon).
- In **1930** - conventional tomography (A. Vallebona).
- In **1950/1963** Allan McLeod Cormack develop the theoretical & mathematical methods used to reconstruct CT images.
- In **1971** – first commercial CT (Godfrey N. Hounsfield)
- In **1972** Godfrey N. Hounsfield and colleagues built the CAT scan machine, taking Cormack's theoretical calculation into a real application (1st generation CT) .
- Hounsfield conceived his idea in **1967** and it was publicly announced in **1972**.
- **In 1975** – 2nd & 3rd generation CT.
- In **1976** – 4th generation CT.



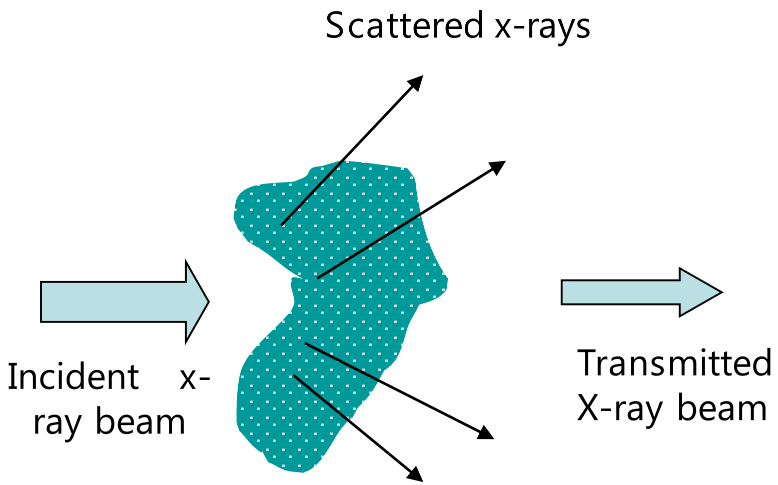
First CT experimental set

Hounsfield

- For their independent efforts, Cormack and Hounsfield shared the Nobel Prize in medicine and physiology in **1979.**
- **In 1984** – 5th generation CT.
- In **1989** – 6th generation CT (single-row CT).
- **In 1994** – double-row spiral CT.
- **In 1998** – 7th generation CT.
- In **2001** – **16** – row spiral CT.
- In **2007** – 320 – row spiral CT.

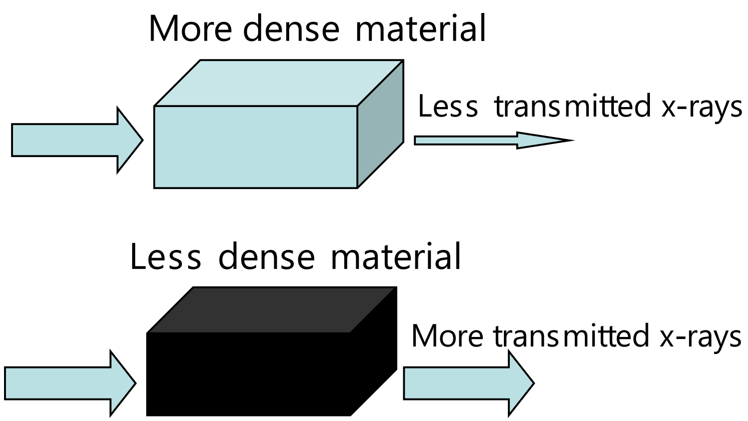
Understanding basic factors:

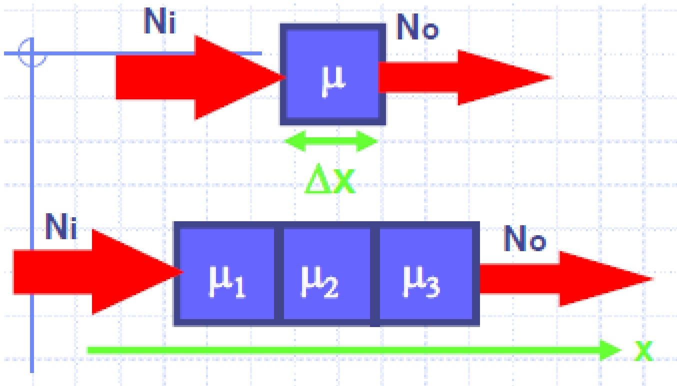
- Absorption: Stopping of X-rays with transfer of energy
- Scatter: Deflection of X-rays
- Incident Intensity: Number of X-ray photons falling on an object
- Transmitted Intensity: Number of photons passing through



Attenuation:

- The reduction of the beam intensity on passing through the material due to absorption plus scatter
- The degree of attenuation is obtained by measuring and comparing the incident and transmitted intensities



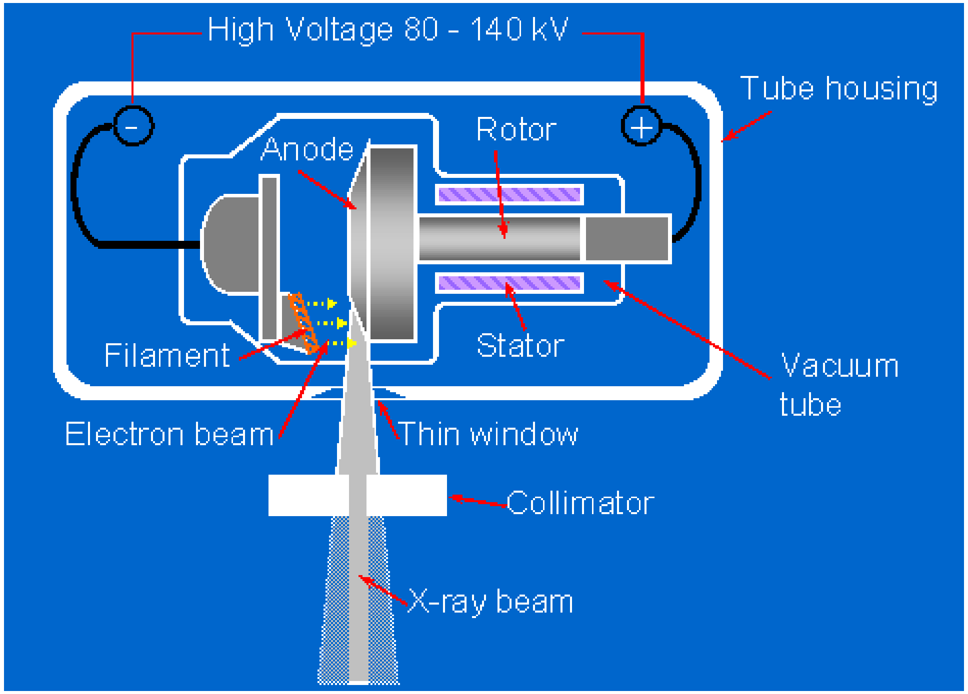


$$N_o = N_i e^{-\mu \Delta x}$$

$$N_o = N_i e^{-(\mu_1 + \mu_2 + \mu_3)\Delta x}$$

X-ray journey:

- X-rays produced by firing electrons at metal target
- X-ray beam filtered to optimise spectrum
- Beam shaping filter conditions quality and intensity of beam
- X-rays interact with patient
- X-rays absorbed in detectors



X-ray Tube:

- Vacuum
  - ✓ Accelerating electrons
  - ✓ Electrons will travel faster
- Filament
  - ✓ Alternating current
  - ✓ Thermal electrons
- Cathode (-)
  - ✓ Filament plate with a tiny slit
  - ✓ Connected to high voltage battery source
- Target/Anode (+)
  - ✓ Electrons collide with target
  - ✓ Produce x-ray
  - ✓ Must have high melting point (Tugnsten)

Applications of X-ray attenuation & detection:

- Conventional X-ray (Radiography)
- Conventional Tomography
- Computed Tomography

1) Conventional X-ray:

- Conventional X-ray produces a compression of a volume to a plane.
- The detector is the silver halide crystal on a X-ray film.
- The degree of blackening represents the total attenuation through the path of X-ray photons.
- The higher the attenuation the lesser is the blackness.
- The structure which results more attenuation or more transmission predominates in the image.

2) Conventional Tomography:

- The source and detector moves in the opposite direction.
- Produces images of coronal or sagittal sections (cuts) of areas of interest.
- Eliminates the superimposition of structures above and below.

3) Computed Tomography:

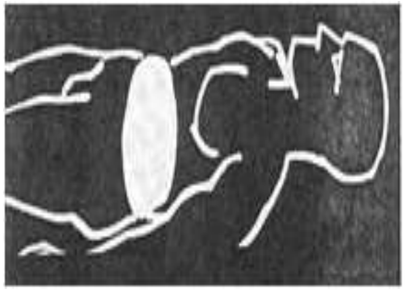
- X-ray beams passed through the object from many points across the object from many angles (projections).
- Opposite the object is an array of detectors that measure the intensity of the X-ray beam at points laterally across the object.
- **Computed tomography (CT) is a medical imaging method employing tomography.**
- The data acquired by the detectors with each slice is electronically stored and are mathematically manipulated to compute across sectional slice of the object.
- **A large series of two-dimensional X-ray images (slices) of the inside of an object are taken around a single axis of rotation.**
- **Digital geometry processing is used to generate 3D images of the object from those slices.**
- 3D information can be obtained by comparing slices taken at different points along the object and the computer can create a 3D image by stacking together slices.

Why CT?

- Conventional radiography suffers from collapsing of 3D structures onto a 2D image.
- Although the resolution of CT is lower, it has extremely good low contrast resolution enabling the detection of very small changes in tissue type.
- CT gives accurate diagnostic information about the distribution of structures inside the body.

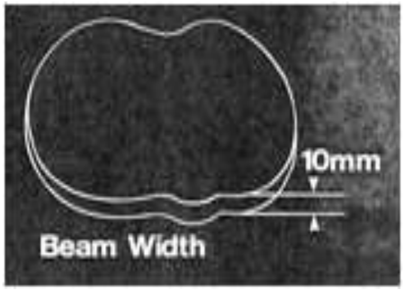
CT scan:

- The CT image is recorded through a SCAN.
- CT scan produces axial sections/cuts/slices
- A scan is made up of multiple X-ray attenuation measurements around an objects periphery.



Section/Cut/Slice:

- The cross sectional portion of the body which is scanned for the production of CT image is called a section/cut/slice.
- The slice has width and therefore volume.
- The width is determined by the width of the X-ray beam.



CT scanning applications:

- Very wide ranging – good for imaging bone and soft tissue
  - ✓ diagnostic imaging
  - ✓ radiotherapy planning: to take X-rays (slices) of the body.
- CT scans take slices and turn into 3D images.

Advantages of CT scanning:

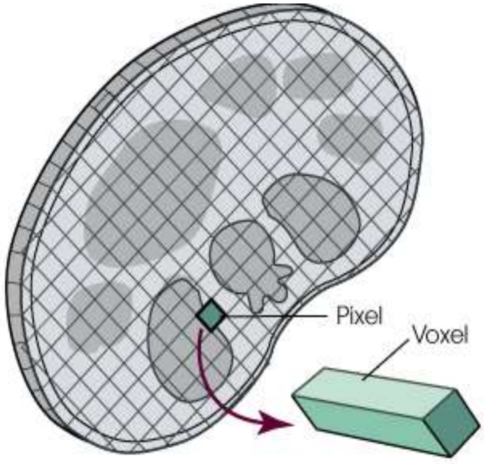
- Ability of differentiate overlying structure.
- Excellent contrast: Overlying structure don't decrease contrast

The Basic CT Term:

- ❑ Image matrix
- ❑ Linear attenuation coefficient
- ❑ CT numbers

Image matrix/Tomographic images:

- The tomographic image is a picture of a slab of the patient's anatomy.
- The 2D array of pixels (picture elements) in the CT image corresponds to an equal number of 3D voxels (volume elements) in the object.
- Every CT slice is subdivided into a matrix of volume element (voxel)
- The viewed image is then reconstructed as a corresponding matrix of picture element (pixel).
- Array of numbers arranged in a grid of rows and columns called a matrix.
- Single square, or picture element, with in the matrix is called a pixel.
- Slice thickness gives the pixel and added dimension called the volume element, or voxel.
- Each pixel is assigned a numerical value (CT number), which is the average of all the attenuation values contained within the corresponding voxel.



Pixel size = FOV/matrix size

Voxel size =
pixel size X slice thickness

➢The diameter of image reconstruction is called the **field of view** (FOV).

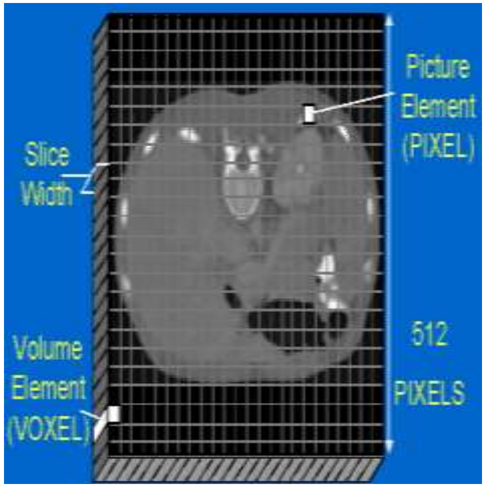
➢ Voxels have the same in plane dimension as pixels, but they also include the slice thickness dimension.

➢ The field of view determines the amount of data to be displayed on the monitor.

➢ On the CRT, each pixel within the image is assigned a level of gray.

➢ The gray level assigned to each pixel corresponds to the CT number or Hounsfield units for that pixel.

➢ Each voxel is about 1mm on a side and is as thick as 2–10mm depending on the depth of the scanning X-ray beam.



Linear attenuation coefficient **(μ):**

✓ Basic property of matter

✓ Depends on X-ray energy and atomic number (Z) of materials.

✓ Attenuation coefficient reflects the degree to which X-ray intensity is reduced by a material/object.

CT Numbers or Hounsfield Units:

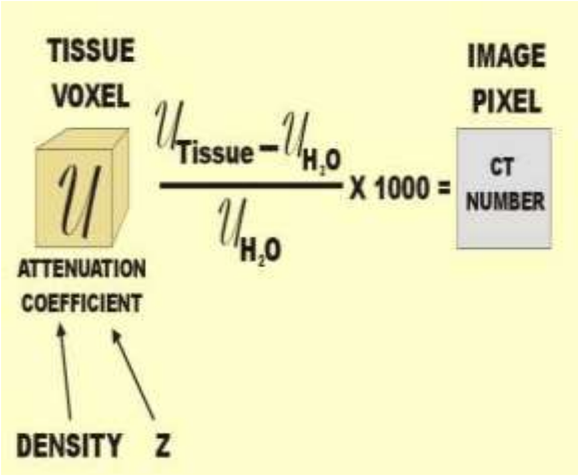
➢ After CT reconstruction, each pixel in the image is represented by a high-precision floating point number that is useful for computation but less useful for display.

✓ Most computer display hardware makes use of integer images.

✓ Consequently, after CT reconstruction, but before storing and displaying, CT images are normalized and truncated to integer values.

❖ CT numbers (or <u>Hounsfield units</u>) represent the percent difference between the X-ray attenuation coefficient for a voxel and that of water multiplied by 1000.

❖ where $\mu(x,y)$ is the floating point number of the *(x,y)* pixel before conversion, μ_{water} is the attenuation coefficient of water, and CT(x,y) is the CT number (or Hounsfield unit) that ends up in the final clinical CT image.



When k=1000, the CT numbers are Hounsfield units (HU)

$$CT(x,y) = 1{,}000\,\frac{\mu(x,y) - \mu_{water}}{\mu_{water}}$$

The precise CT number of any given pixel is calculated from the X-ray attenuation coefficient of the tissue contained in the voxel.

- ➢ CT number ranged from -1000~3095 (12 bit)
- ➢ CT numbers normalized in this manner provide a range of several CT numbers for 1% change in attenuation coefficient.
- ➢ Gray levels on CT image represent attenuation in each pixel
- ➢ Gray levels expressed in Hounsfield units (HU)

❖ Water has a CT number of zero HU and the numbers can be positive or negative depending on the absorption coefficient.

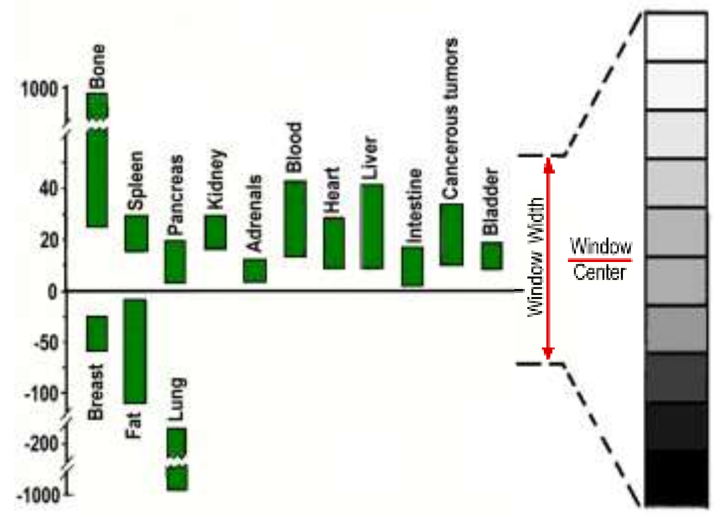
❖ The value of μ_{water} is about 0.195 for the X-ray beam energies typically used in CT scanning

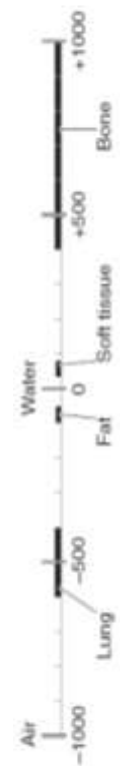
❖ Air is -1000 HU

❖ Bone is 1000 – 3000 HU

✓ **HU represents the linear attenuation of a material**

✓ This normalization results in CT numbers ranging from about - 1,000 to +3,000, where -1,000 corresponds to air, soft tissues range from -300 to -100, water is 0, and dense bone and areas filled with contrast agent range up to +3,000.



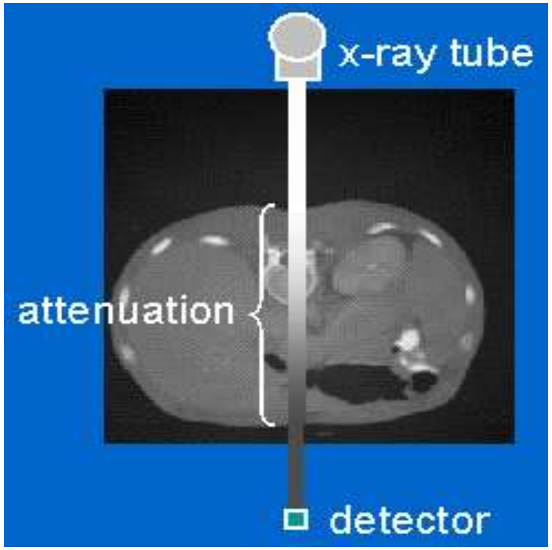


Bone	+400 → +1000
Soft tissue	+40 → +80
Water	0
Fat	−60 → −100
Lung	−400 → −600
Air	−1000

The hounsfield scale of CT numbers

What are we measuring in CT?

- The linear attenuation coefficient (μ), between the X-ray tube and the detector
- The linear attenuation coefficient is a measure of how rapidly are X-ray attenuated



Hounsfield units:

- Each pixel within the matrix is assigned a number that is related to the linear attenuation coefficient of the tissue within each voxel:

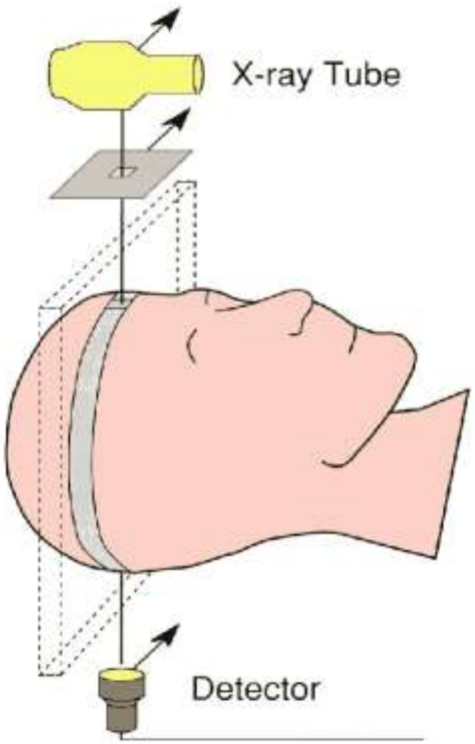
$$I = I_0 \exp(-\mu x)$$

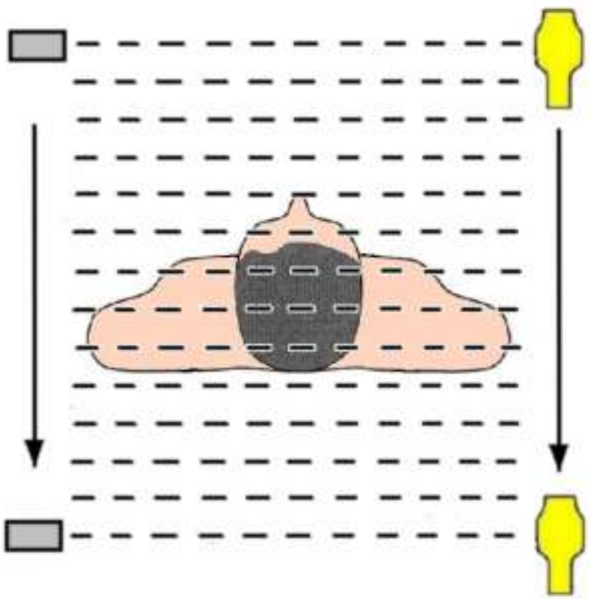
- These are CT numbers or Hounsfield units.
  Defined: A relative comparison of X-ray attenuation of a voxel of tissue to an equal volume of water.

Tomographic acquisition:

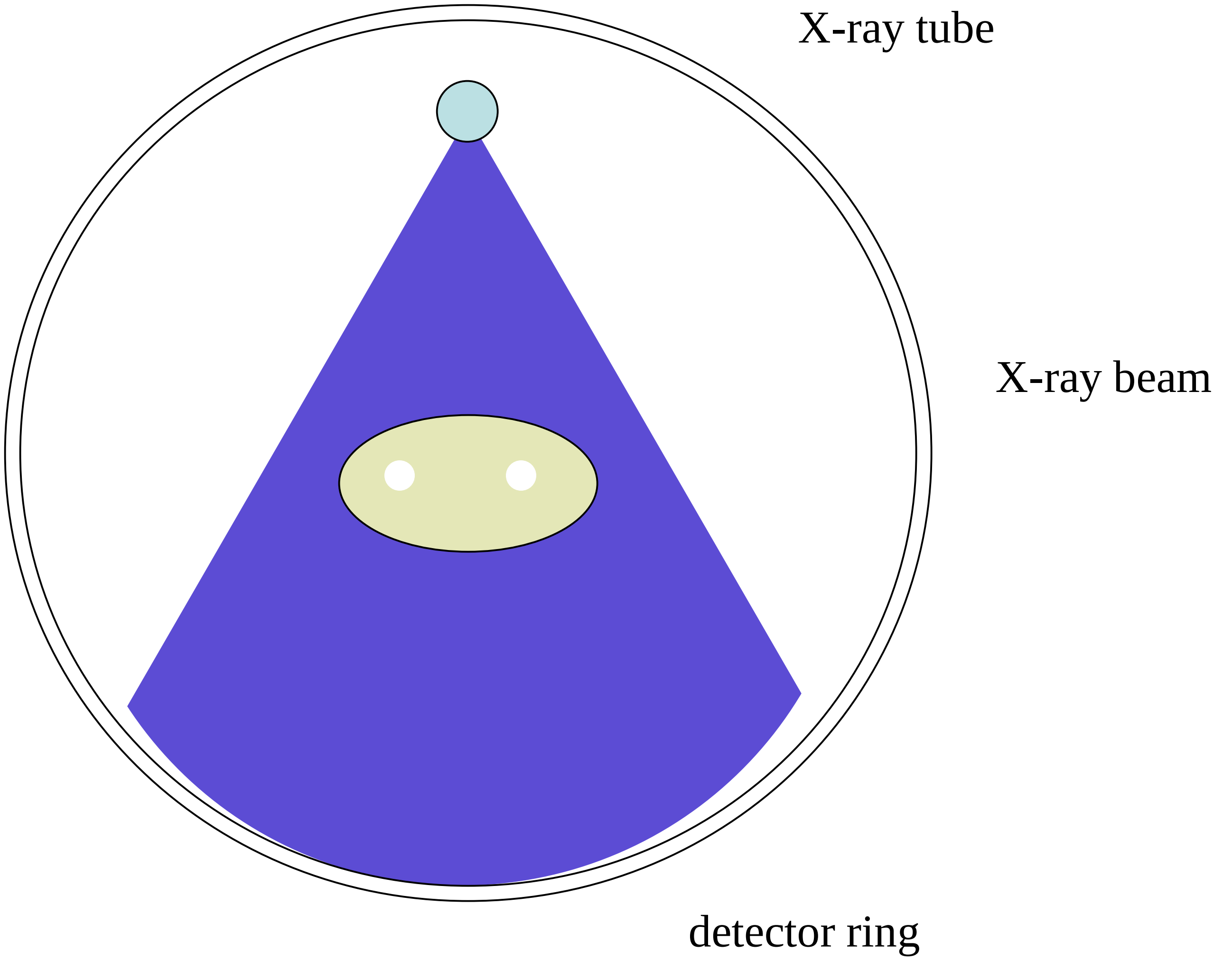
- CT images are produced from a large number of X-ray transmission measurement through the object made by a single detector at a given moment in time is called a ray.
- A series of rays that pass through the object at the same orientation/geometries is called a projection or view.
- Two projection geometries have been used in CT imaging:
- Parallel beam geometry: all rays in a projection parallel to one another.
- Fan beam geometry: rays at a given projection angle diverge.
- All modern CT scanners incorporate fan beam geometry in the acquisition and reconstruction process.
- Purpose of CT scanner hardware is to acquire a large number of transmission measurements through the object at different position is called acquisition.

The basics of image formation - Parallel beam geometry



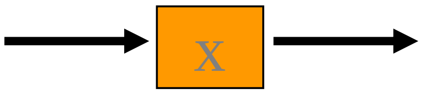


The basics of image formation - Fan beam geometry:



Tomographic reconstruction:

- Each X-ray acquired in CT is a transmission measurement through the object along a line, detector measures an X-ray intensity (I_t).
- The un attenuated intensity of the X-ray beam is also measured during the scan by a reference detector (I_o).
- The relationship between I_t and I_o is:

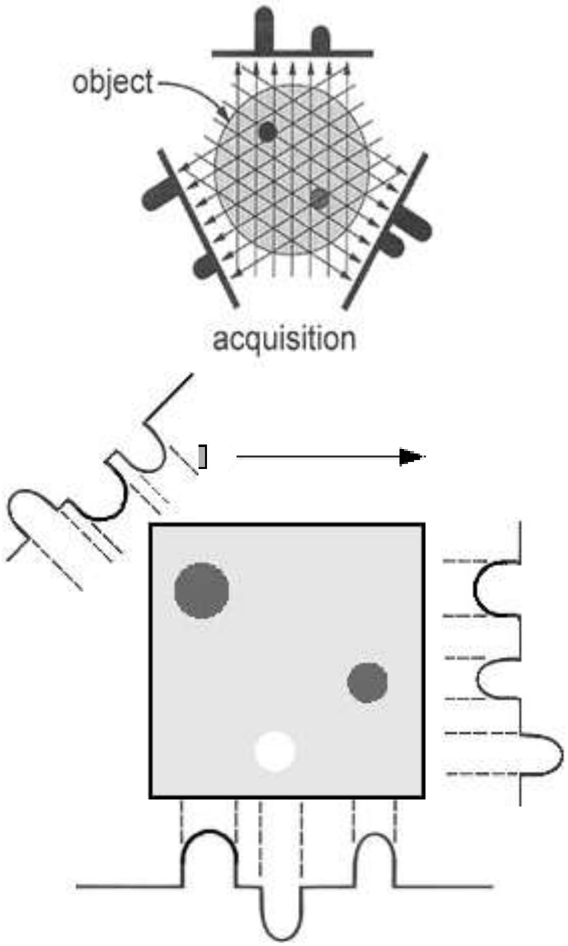


$$I_t = I_0 e^{-\mu t}$$

$$\ln(I_0 / I_t) = \mu t$$

  where t is the thickness of the X-ray, μ is the average linear attenuation coefficient along the ray.
- Filtered backprojection reconstruction method is most widely used in clinical CT scanners.
- This method builds up the CT image by essentially reversing the acquisition steps.
- During backprojection reconstruction, the μ value for each ray is measured along this same path in the image of the patient.
- As data from a large number of rays are backprojected onto the image matrix, areas of high attenuation tend to reinforce one another, as do areas of low attenuation, building up the image in the computer.

Acquisition steps



- CT image from the acquisition data essentially reverses the acquisition geometry mathematically.
- Each transmission measurement is backprojected onto a digital matrix.
- After backprojection areas of high attenuation are positively reinforced through the backprojection process.
- Whereas other areas are not and thus the image is built up from the large collection of rays passing through it.

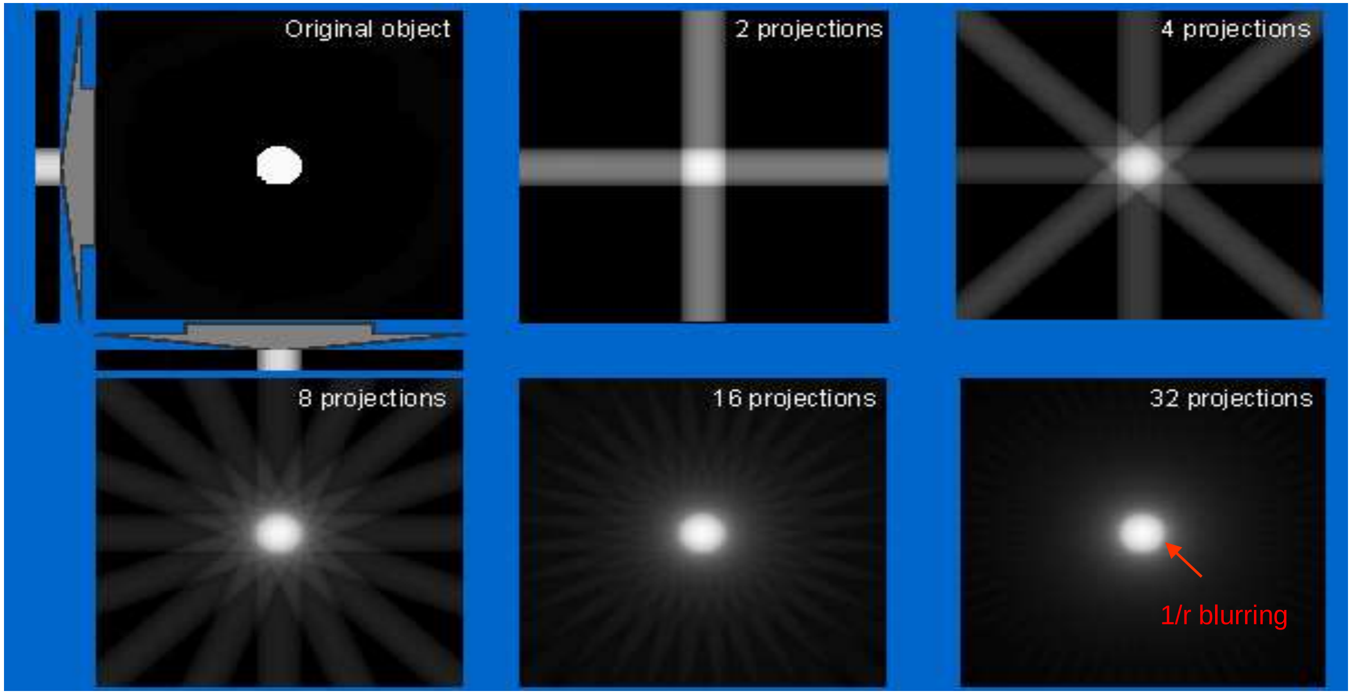
Reconstruction algorithms:

Computer based:

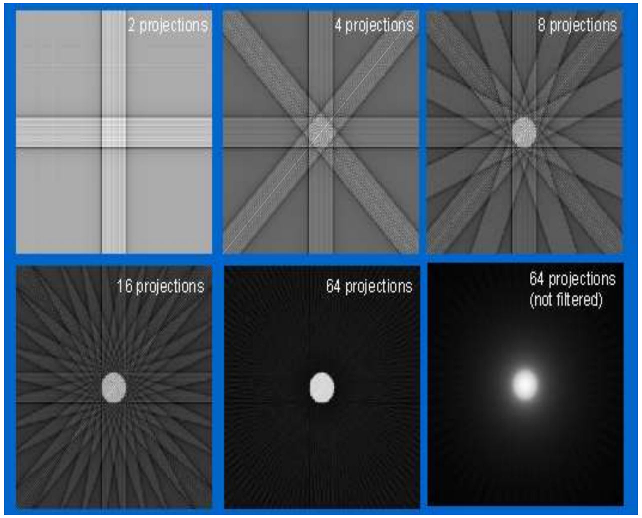
- ❖ Simple back-projection
- ❖ Filtered back-projection
- ❖ Iterative techniques

Simple back-projection: It's produces blurred images.

- This projection data need to be filtered before reconstruction.
  - ❖ Reverse the process of measurement of projection data to reconstruct an image.
  - ❖ Each projection is uniformly distributed across the reconstructed image.



Filtered back-projection:



- Simple back-projection produces blurred images
- Projection data need to be filtered before reconstruction is called **Filtered Backprojection**.
- Filtered backprojection removes the star-like blurring seen in simple backprojection
- Different filters can be applied for different diagnostic purposes

    - Smoother filters for viewing soft tissue
    - Sharp filters for high resolution images

- "Filtered" refers to use digital algorithms called convolution to improve image quality or change certain image quality characteristics, such as detail and noise
- Back-projection is the same as before

Filter:

- a de-blurring function is combined (convolved) with the projection data to remove most of the blurring before the data are backprojected.
- A high-frequency filter reduces noise and makes the image appear "smoother."
- A low-frequency filter enhances edges and makes the image "shaper."
- A low-frequency filter may be referred to as a "high-pass" filter because it suppresses low frequencies and allows high frequencies to pass.

The filtered backprojection process involves the following steps:

- Generating a sinogram from a set of N projections.
- Filtering the data to compensate for blurring
- Backprojecting the data
- Projection and sinogram

✓ Ray: the X-ray read by every one detector within a short time interval.
✓ Projection: all rays sum in a direction
✓ Sinogram: all projections

- A five-stage numerical example for filtered backprojection as shown in figure.
- This having a high central value of 7 surrounded by lower values of 1.
- Moving on to the first computation: the horizontal back-projection in (b) stores the total ray sum in each row of (a).
- In the second stage (c) right diagonal ray sums are then added followed by the vertical (d) and left diagonal ray sums, giving the final image in figure (e).
- Different filters can be applied for different diagnostic purposes:
  - smoother filters for viewing soft tissue
  - sharp filters for high resolution images

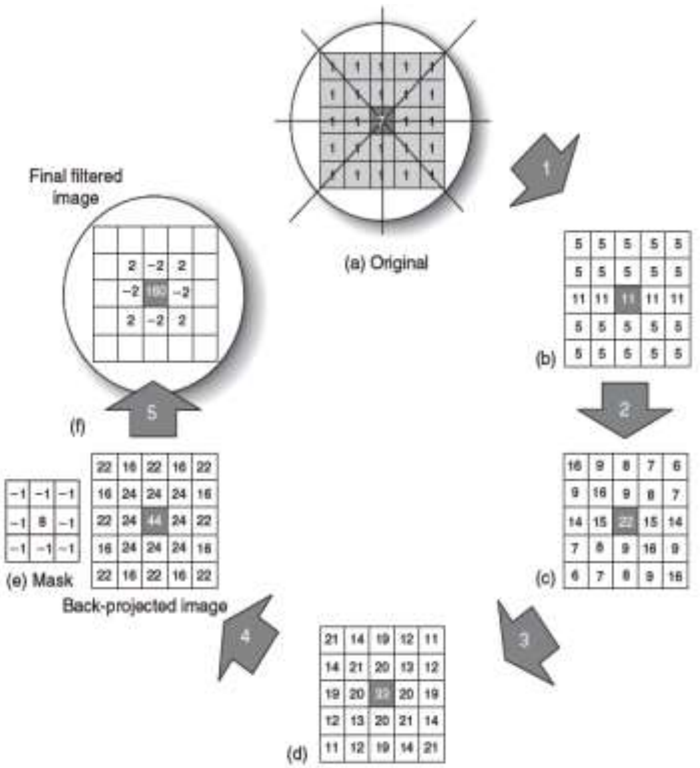
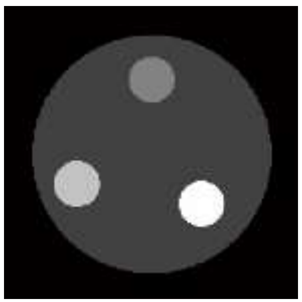
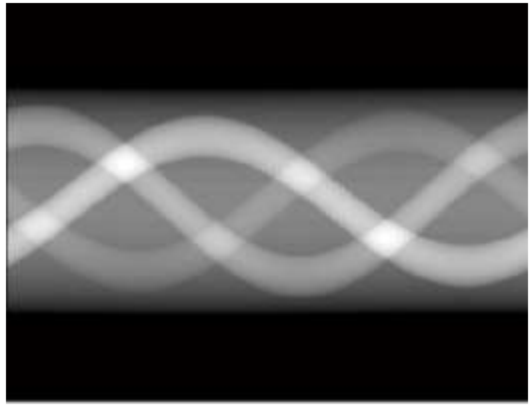
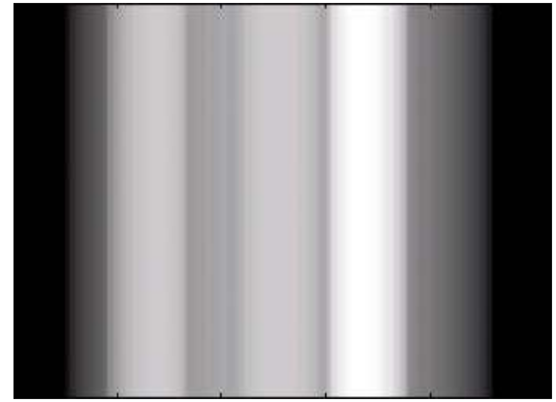


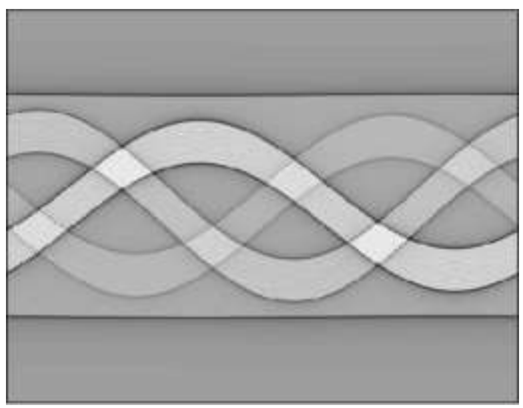
Image reconstruction:

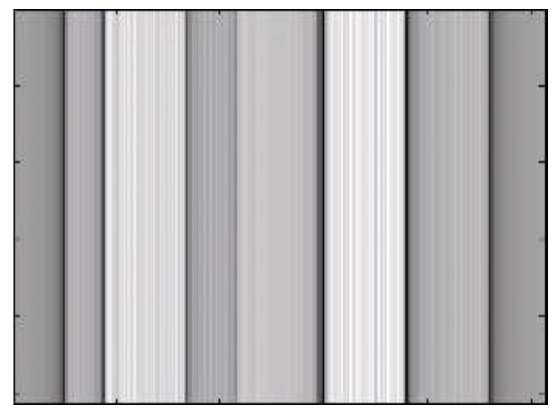






Simple back-projection

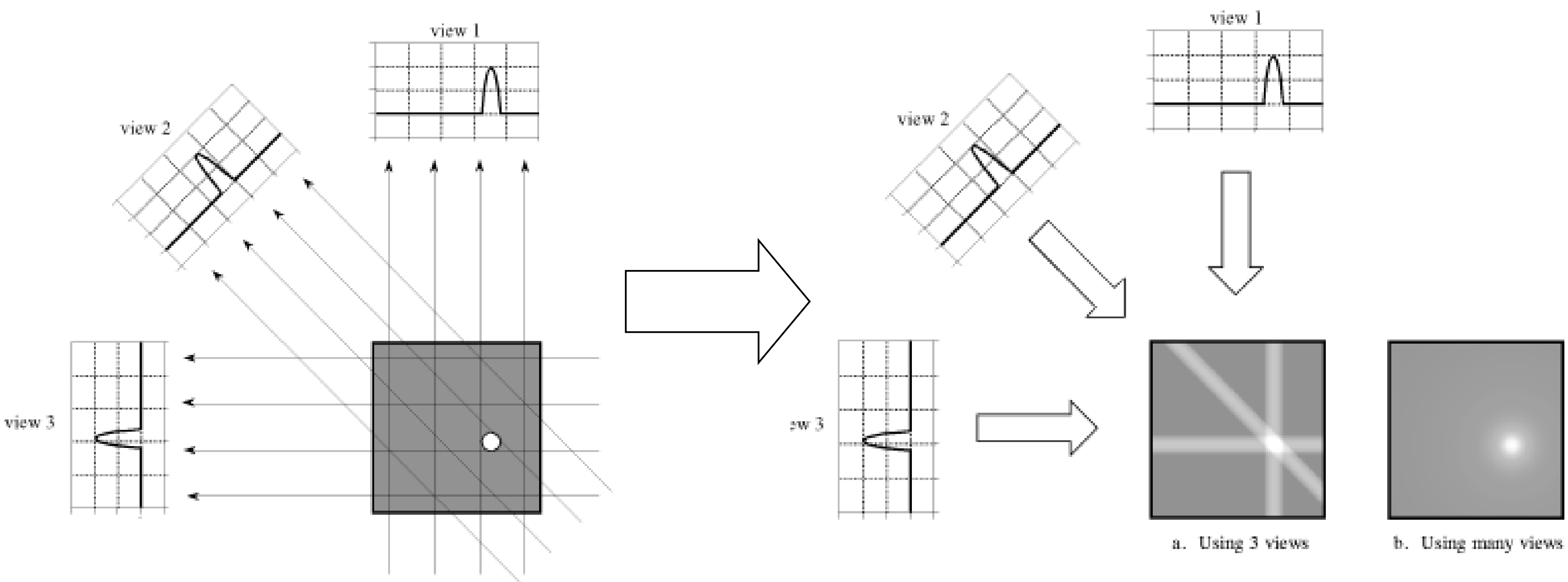




Filtered back-projection

Backprojection:

- Backprojection is the actual process used to produce or "reconstruct" the image.
- A tightly collimated X-ray beam is used and its total absorption provides the ray-sum signal for each matrix row, which is stored in the array processor image memory.
- The central high uptake, from the original, is now distinguishable but it has a star-burst interference pattern.
  ✓Backprojection data (in Sinogram) → 1D-FT → filled in k-space → central slice projection theorem → 2D-inverse FT → CT image

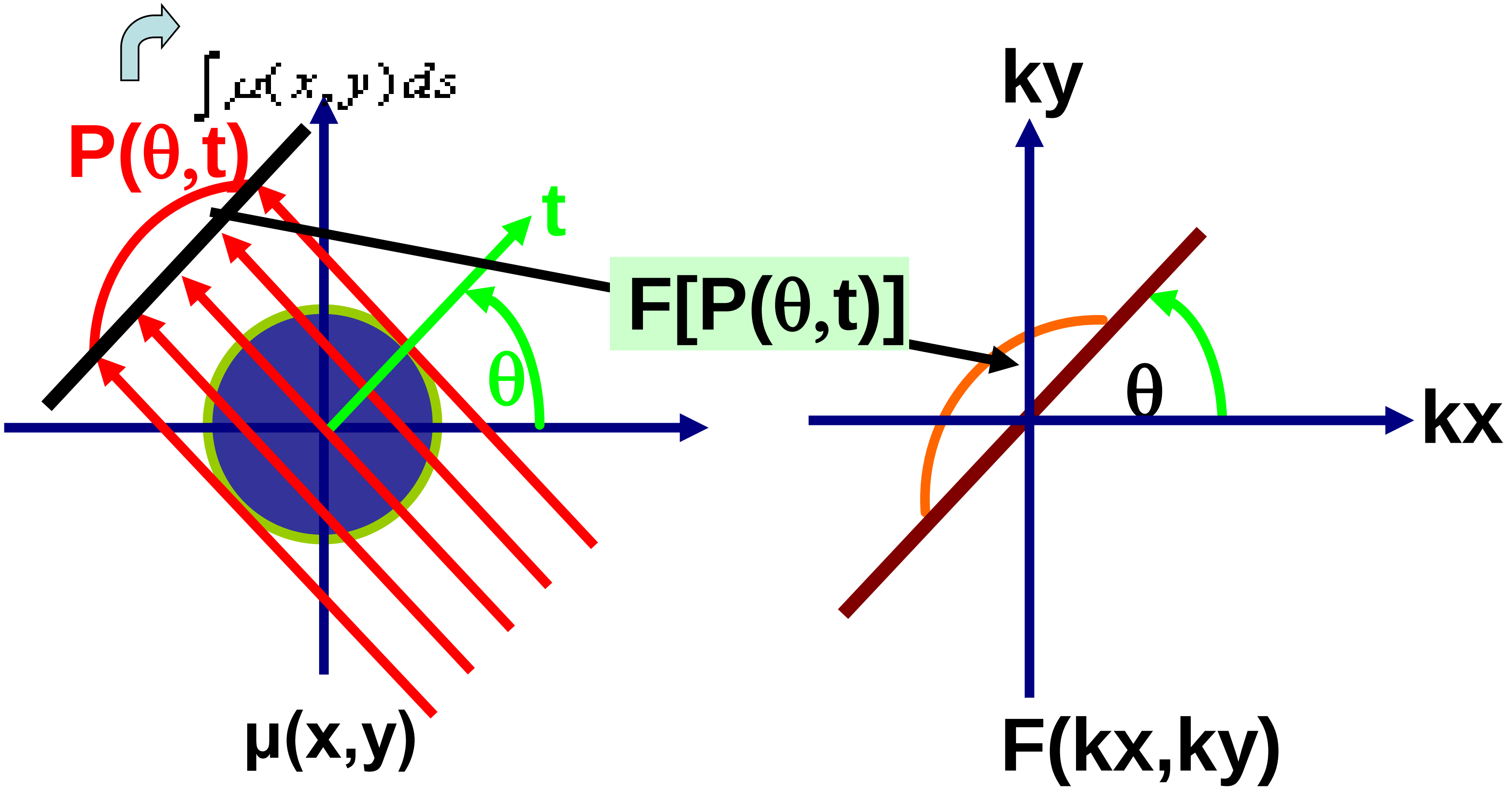


Convolution:

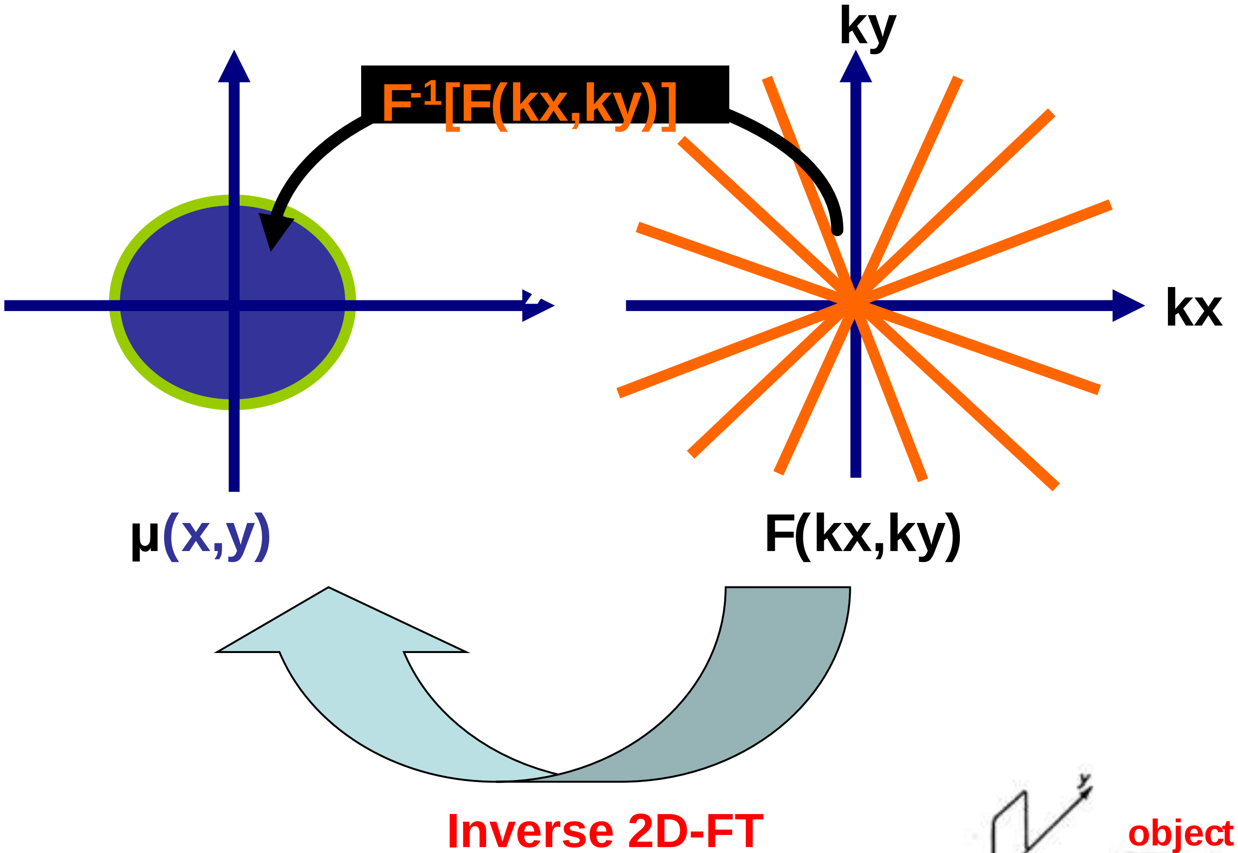
- The back-projected matrix is subjected to a filter mask, or convolution kernel whose contents are multiplied with the back-projected image.
- The filter mask consists of a small symmetrical matrix containing a set of numbers. These can have positive or negative values.

Central Slice Projection Theorem (CSPT):

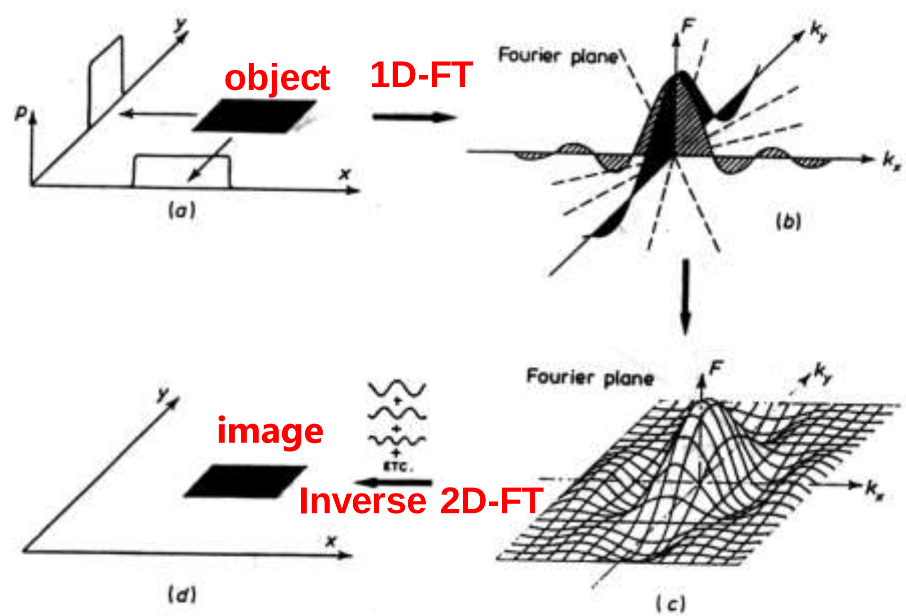
If a 1D Fourier Transform is performed on a projection of an object of some angle, the result will be identical to one line on 2D FT of that object and at that angle.



CSPT can relate the fourier transform of the projection to one line in the 2D K space formed by the 2D Fourier transform of **μ** (x,y)



ky
$F^{-1}[F(kx,ky)]$
kx
μ(x,y)
F(kx,ky)
Inverse 2D-FT



object
1D-FT
Fourier plane
F
k_y
k_x
p
y
x
(a)
(b)
image
Inverse 2D-FT
ETC.
(c)
(d)

A simple example of reconstruction based on data from 4 angles (projection) :

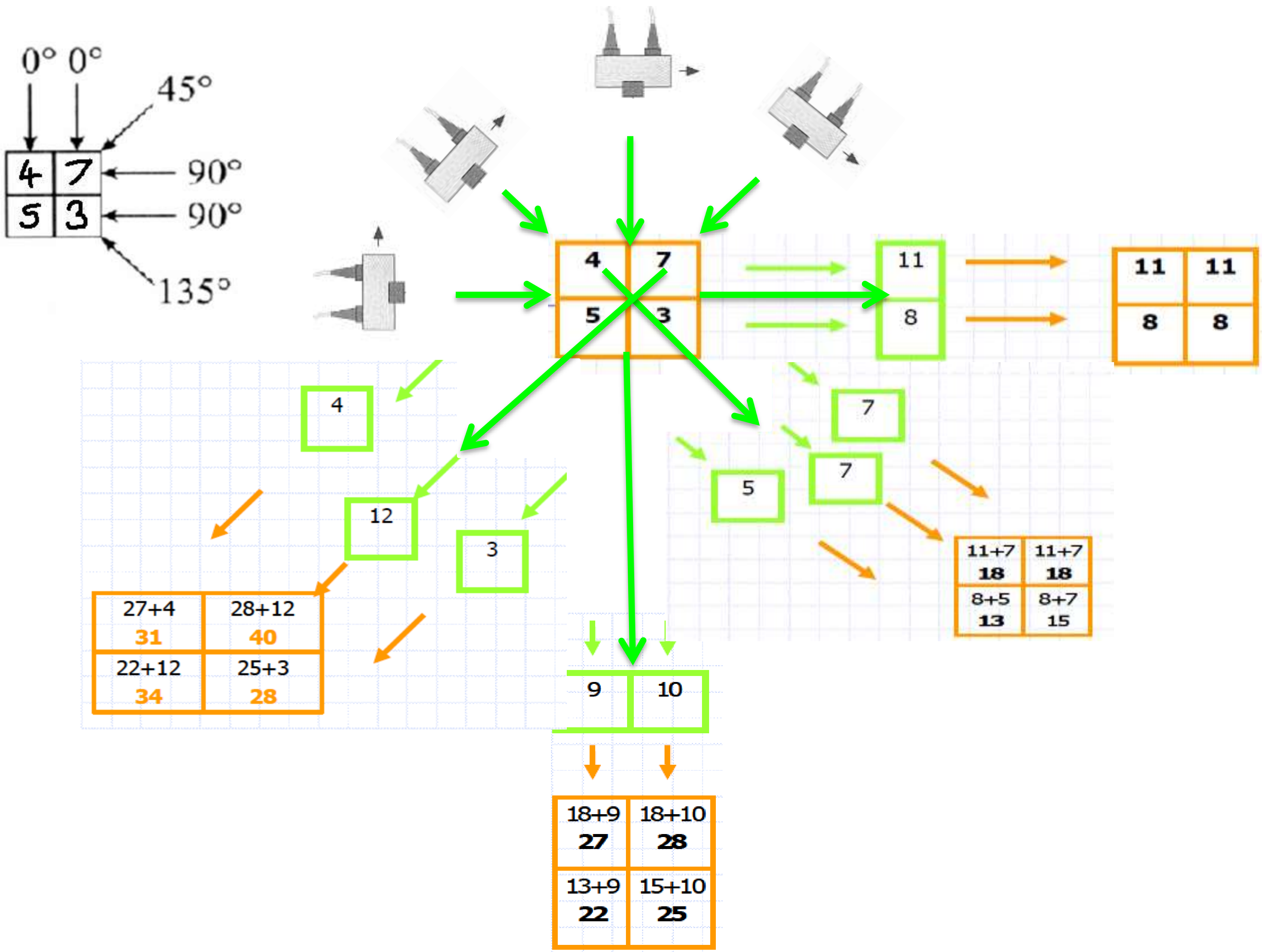


Image data post-processing:

- After the initial processing of the acquired image data (amplification and log transformation) and before producing the clinical image data is subjected to a series of post-processing measures that are used to improve image quality. These are:
  - Windowing
  - Spatial filtering
  - Temporal or recursive filtering
- Edge enhancement, windowing, and magnification are a few of the image post-processing features that can improve the diagnostic quality of the images.
- Quantitative analysis programs are also available in the form of ejection fraction calculation and wall motion measurements.

Windowing:

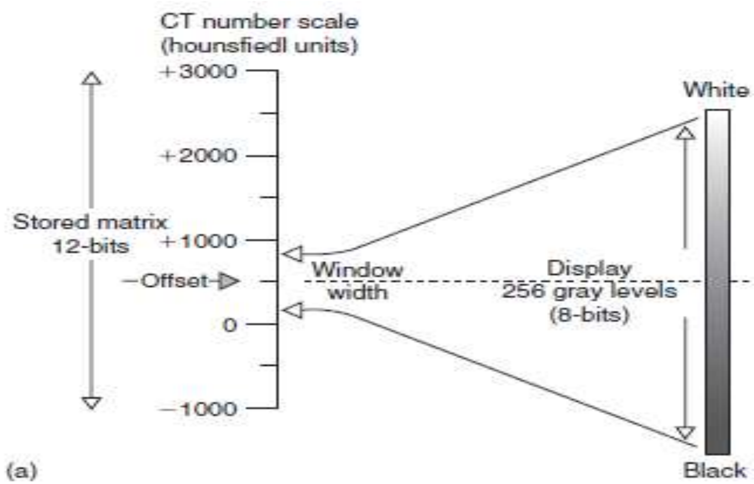
- Postprocessing procedure is called windowing and allows small contrast changes to be amplified.
- Windowing a section of the stored data allows a selected anatomy to be viewed with improved contrast visibility in CT.
- A window permits a small section of these stored values to be displayed using the full 8 bits or a reduced bit number if greater contrast between structures is required.
- By varying the position and width of the window it is possible to display a section of the total range on a black to white scale.
- The center and width of the window may be chosen independently.
- It is therefore possible to represent any gray level range of the digitized signal with maximum contrast resolution on the viewing monitor.
- The maximum window possible is all 4096 gray levels displayed simultaneously as black to white.
- The minimum range occurs when one level is set to black and the next level up is set to white. Thus the maximum contrast between adjacent levels is obtained.

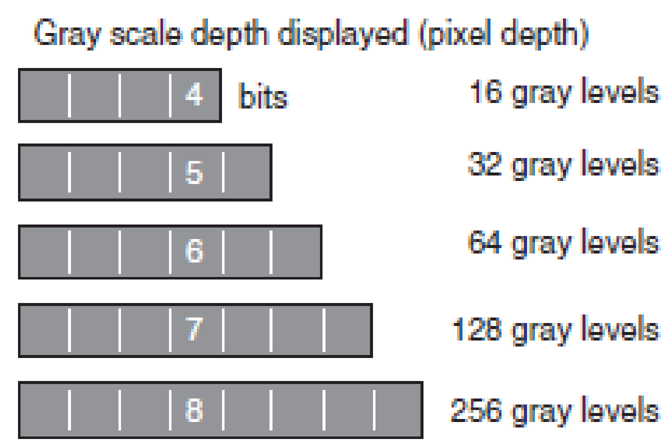
Windowing and image display:

Windowing: Displaying a selected range of CT numbers stored in a voxel array.

Iimage display:

- Under ideal conditions the eye can distinguish between 50 and 80 gray levels on a good quality computer monitor.
- Routinely this range is reduced to nearer 35 to 40.
- Since the full CT data set of 4000 levels cannot be displayed at one time the user must select, or **window**, a range of CT values for display.
- The CT number range: a window width has been selected with a particular offset, which is the central value.
- The **window width** (4 to 8 bits) can be displayed at various pixel depths are used for sampling the available range of CT numbers.
- The position of the window within the data is the **offset**.

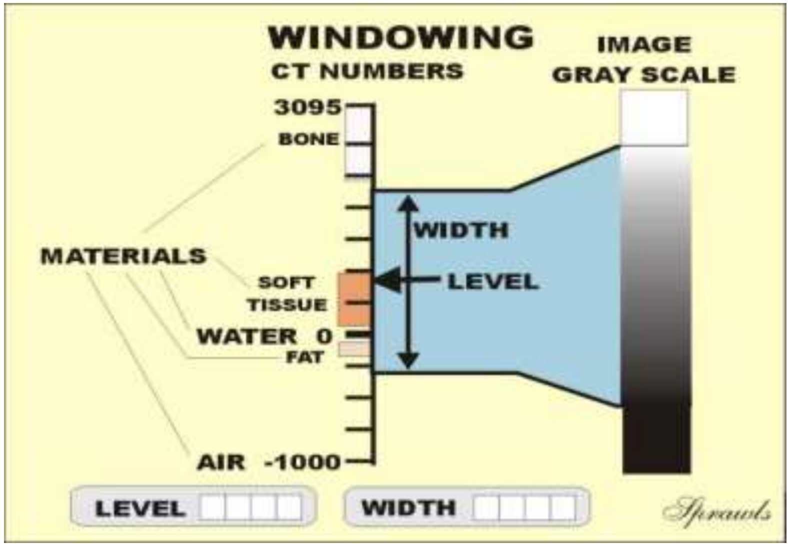




- If a broad window, from 1000 to 1000 is displayed using 256 gray levels, a CT number difference of 8 is represented by one gray level, giving a poor contrast image. Such an image would be useful as an overview only.
- Small changes in CT number, and therefore greater contrast, can be obtained from narrower windows.
- If it is too small the image can show a great deal of noise and details in bony structures or fatty tissue could be missed.
- The window itself can be moved or offset up or down the scale to include soft tissue or bone detail.
- CT values outside the window will be shown as white (above the window value) or black (below the window value).
- For the differentiation of bony or soft tissues certain window offsets are recommended.
- For most soft tissue images a window offset of between 35 to 40 and a window width of 200 to 450 covers most detail.
- For thorax–lung the offset should be 700 and for the inner ear 200.
- Double windowing is available on some machines where two different density ranges can be displayed together, i.e. negative lung tissue alongside positive detail of the mediastinum.
- An entire gray scale is available for each of the window widths and, for clarity, a bright contour separates them.

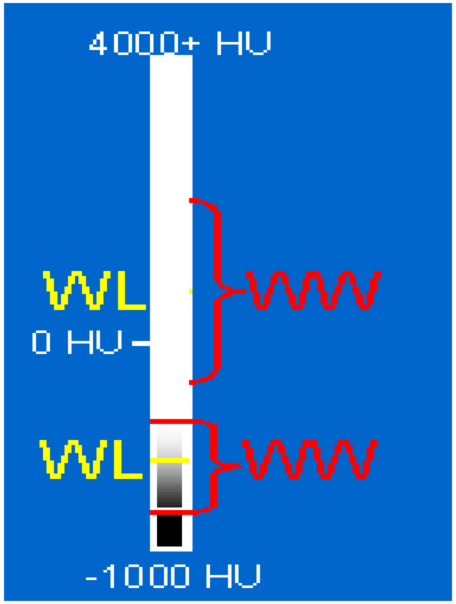
Window width (WW) and Window level (WL) :

- The window width covers CT numbers of all the tissue of interest that is displayed as shades of gray, ranging from black to white. Thus width controls the contrast in the displayed image.
- The level control adjust the center of the window and identifies the type of tissue to be imaged.



Reducing window width increases the displayed image contrast among the tissues

- This technique allows the technologist to alter the contrast of the displayed image
- ✓ by adjusting the window width (WW) and window level (WL).
- ❖ Window width: is the range of CT numbers that are used to map signals into shades of gray (HU from black to white).
- Wide/Narrow or Long/Short
- ❖ Window level: determines the midpoint of the range of gray levels to be displayed
- Darker or Lighter



CT number window:

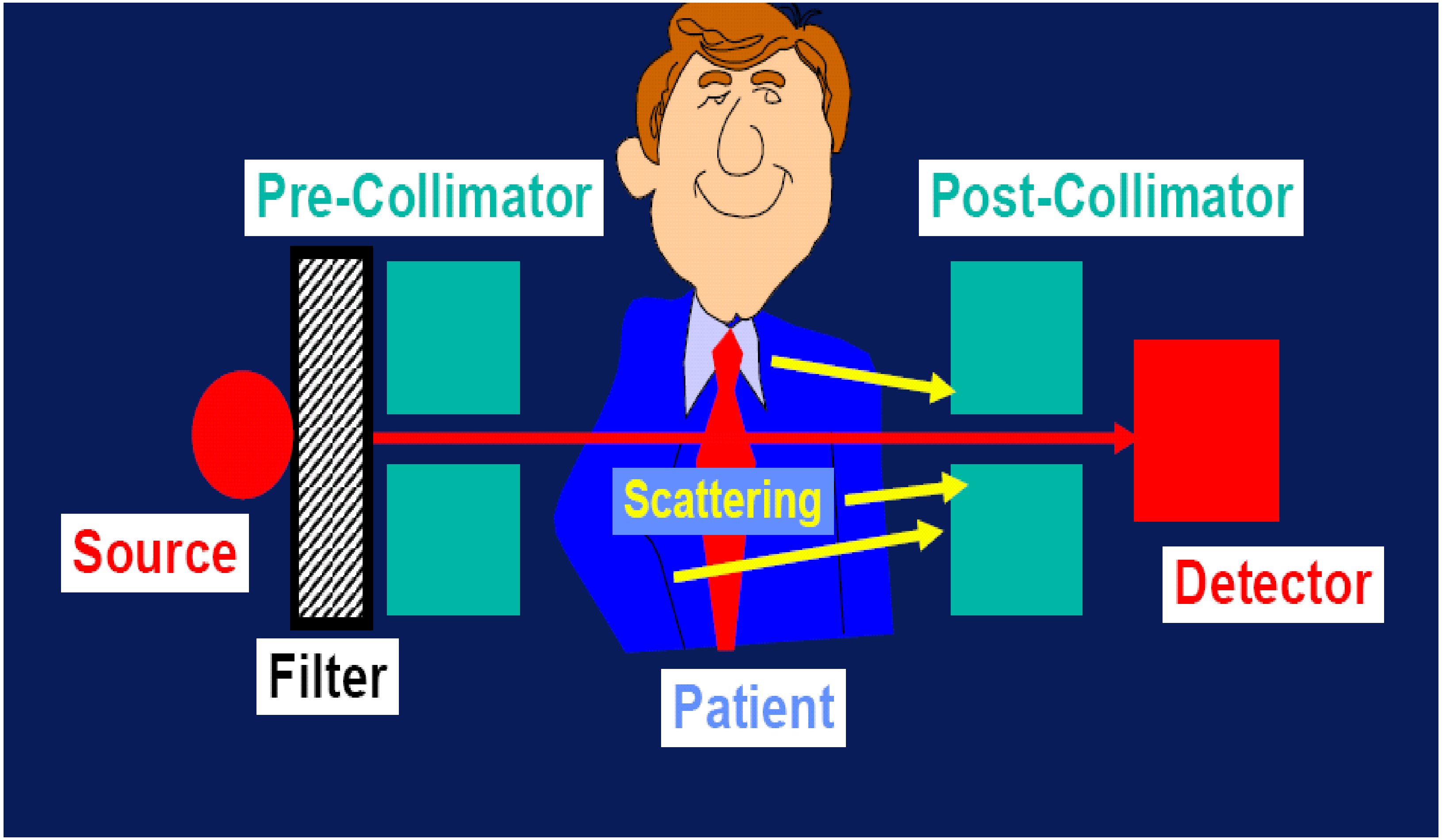
- ➢ CT images can be displayed with arbitrary brightness and contrast
- ➢ Most common: windowing or gray-level mapping
- ➢ Display is defined using window level (WL) and window width (WW)
- ➢ WL is CT number of mid – gray
- ➢ WW is number of HU from black – white
- ➢ Same image data at different WL and WW

CT for radiotherapy – calibration, HU to mass density conversion:

- ❖ HU do not represent mass density, needed for dose calculation, directly.
- ❖ To obtain mass densities of each voxel:
- ❖ A set of tissue equivalent materials with known mass densities is scanned and a calibration curve is created

Collimators:

- Pre-patient collimator- control slice thickness
- Pre-detector collimator-reduce scattered radiation

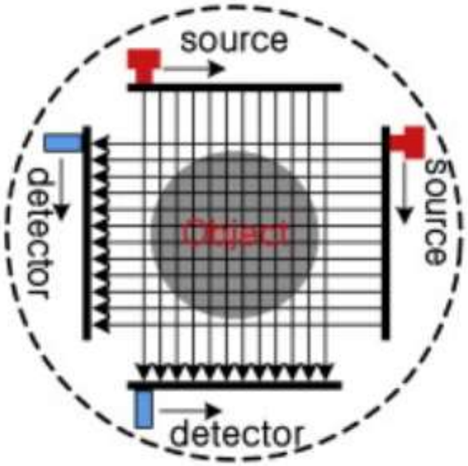


Definition of Generation:

- Classification of computed tomography (CT) based upon:
  - ✓ Arrangement of components (X-ray tube and Detector movement)
  - ✓ Mechanical motion required to collect data
- "Generation" the order in which CT scanner design has been introduced and each has a number associated with it.

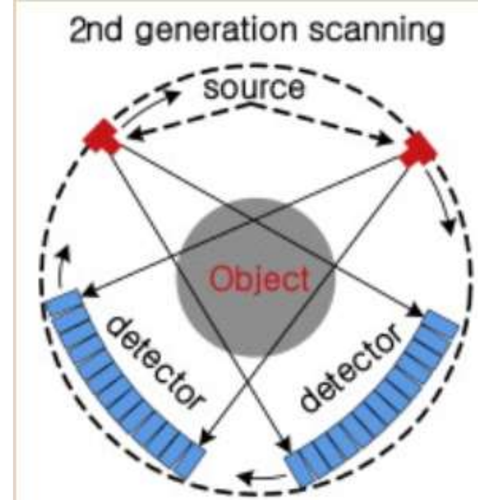
1st **generation (1972)**: rotate/translate, pencil beam:

Design: X-ray tube and single detector are connected and move together by translation and then rotation.



- Source and detector, rigidly coupled
- X-ray beam has liner (pencil-like) shape
  - ✓ translated across patient to obtain set of parallel projection measurements at one angle
- Pencil beam/parallel ray geometry allowed very efficient scatter reduction, best of all scanner generations
- Source/detector rotate slightly and a subsequent set of measurements are obtained during a translation past patient
- Detector cell to collect all the data for a single slice
- Process is repeated once for each projection angle until **180** projections, across a **24** cm FOV
- Translation and rotation process, this geometry is referred to as a translate/rotate scanner
- Contrast resolution of internal structures was unprecedented, images had poor spatial
- Resolution very poor
- NaI detector signal decayed slowly, affecting measurements made temporally too close together

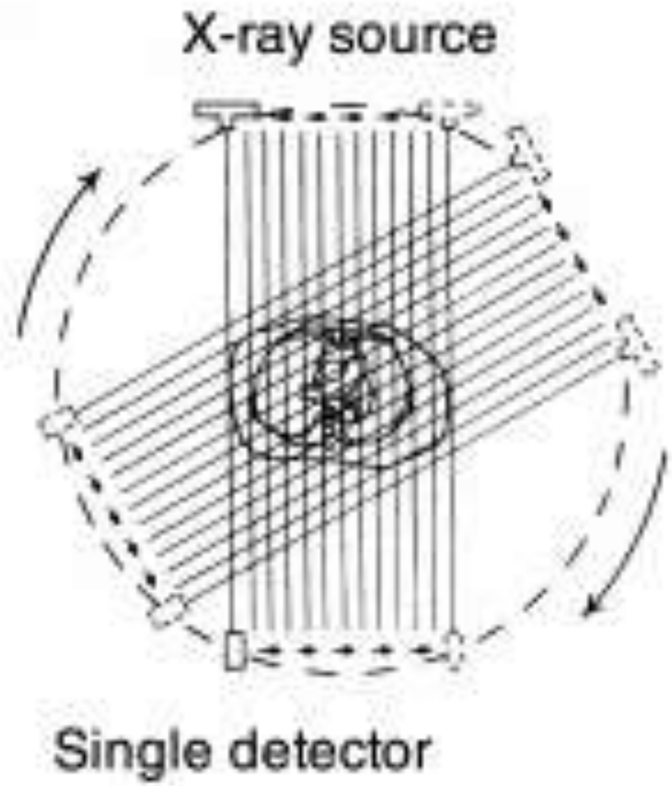
2nd generation (1975): rotate/translate, narrow fan beam



Design: Single X-ray tube and multiple detectors elements arranged in a row

- ✓ X-ray source emits radiation over a large angle (~10°)
- ✓ Narrow fan beam allows more scattered radiation to be detected
- ✓ Incorporated linear array of 30 detectors
- ✓ More data acquired to improve image quality
- ✓ The efficiency of measuring projections was greatly improved (one minute per slice)
- ✓ Shortest scan time was 18 seconds/slice

- ➢ Source and array of detectors are translated as in a first generation system
  - ✓ but since beam measured by each detector is at a slightly different angle with respect to object, each (single) translation step generates multiple parallel ray projections
- ➢ Multiple projections obtained during each traversal past the patient
  - ✓ This scanner is significantly more efficient and faster than 1st one
- ➢ This generation: a translate/rotate scanner
- ➢ Early versions: 3 detectors each displaced by 1°
  - ✓ Since each detector viewed the X-ray tube at a different angle, a single translation produced 3 projections
  - ✓ The system could rotate 3° to the next projection rather than 1°
  - ✓ Make only 60 translations instead of 180 to acquire a complete section
  - ✓ Scan times were reduced

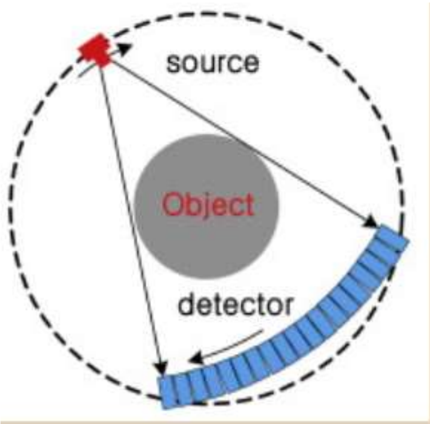


1^{st} generation CT scanner
(Parallel beam,
translate-rotate)

2^{nd} generation CT scanner
(Fan beam, translate-rotate)

3rd generation **(1975)**: rotate/rotate, wide fan beam

Design: Full rotating **(0.5 sec)** movement of single X-ray tube + multiple detectors complex **(300-700** detectors, usually circular**)**

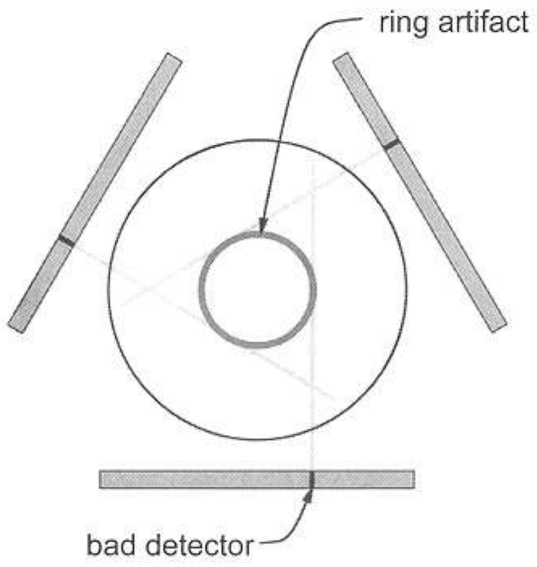


- X-ray beam
  - ✓ X-ray tube is collimated to a wide X-ray beam (fan shaped)
  - ✓ The fan beam is covering all the sample **(~30°)**
  - ✓ Higher-power, rotating anode X-ray tubes, improve scan speeds **(2** sec/slice**)** to cover entire patient
  - ✓ Eliminated need for translational motion, ie., no translation required
  - ✓ Complete **360°** rotation
  - ✓ Rotate/Rotate movement
  - ✓ One rotation = one slice
  - ✓ Much faster **(**few seconds**)**

- Directed toward an arc-shaped row of detectors
- Improvement in detector and data acquisition technology
- Second data acquisition could be made as the tube and detectors move in the opposite direction.
- Current helical scanners are based on modifications of rotate-rotate designs
- Scan times = few seconds or less
- Imaging process is significantly faster than **1**st or **2**nd generation systems
- Reconstruction time ~ seconds

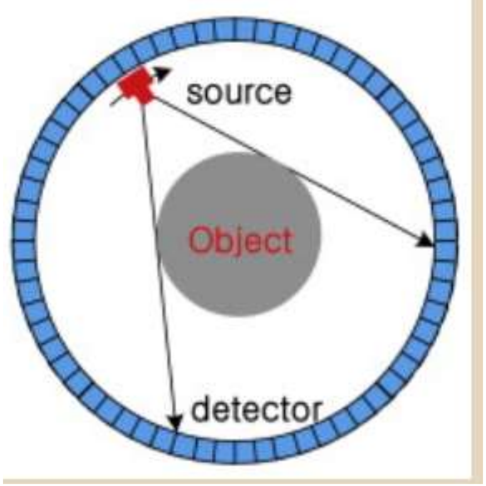
Ring artifacts:

- The rotate/rotate geometry of **3**rd generation scanners leads to a situation in which each detector is responsible for the data corresponding to a ring in the image
- Drift in the signal levels of the detectors over time affects the μt values that are backprojected to produce the CT image, causing ring artifacts



4th generation (1976): rotate/stationary

Design: Only X-ray tube (single) rotates at **360°**, detectors are stationary **(**circular array of FIXED detectors)

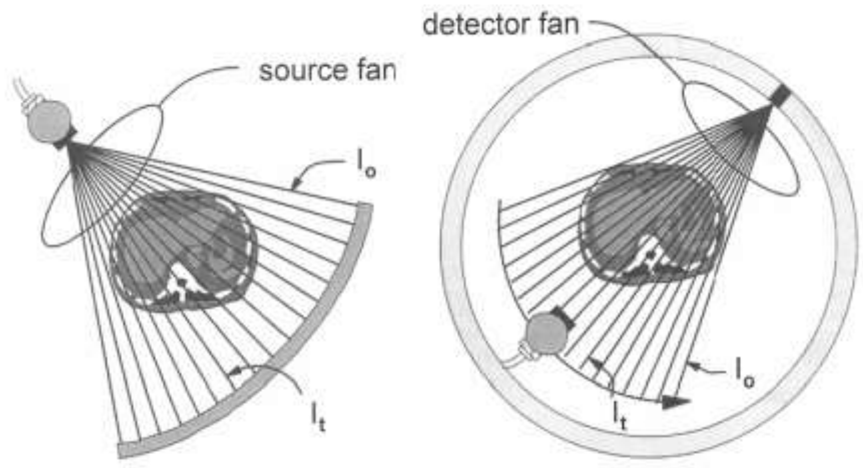


- Larger fan beam
- Rotating X-ray tube provides short bursts of radiation, ie., shorter scanning time
- Detectors collect the remnant radiation to reconstruct into an image

- Fixed ring of as many detectors inside of gantry, completely surrounding the patient, Rotate only movement
- Eliminated translate-rotate motion
- Stationary detector requires a larger acceptance angle for radiation, and is therefore more sensitive to scattered radiation than the **3**rd generation geometry
- Require larger number of detector cells and electronic channels (higher cost) to achieve the same spatial resolution and dose efficiency as a **3**rd generation system
- Designed to overcome the problem of ring artifacts
- Avoids ring artifact problems of **3**rd generation scanners
- Limitation: less efficient use of detectors, less than **1/4** are used at any point during scanning

3^{rd} vs. 4^{th} generation:

- **3^{rd}** generation fan beam geometry has the X-ray tube as the apex of the fan;
- **4^{th}** generation has the individual detector as the apex



third generation

fourth generation

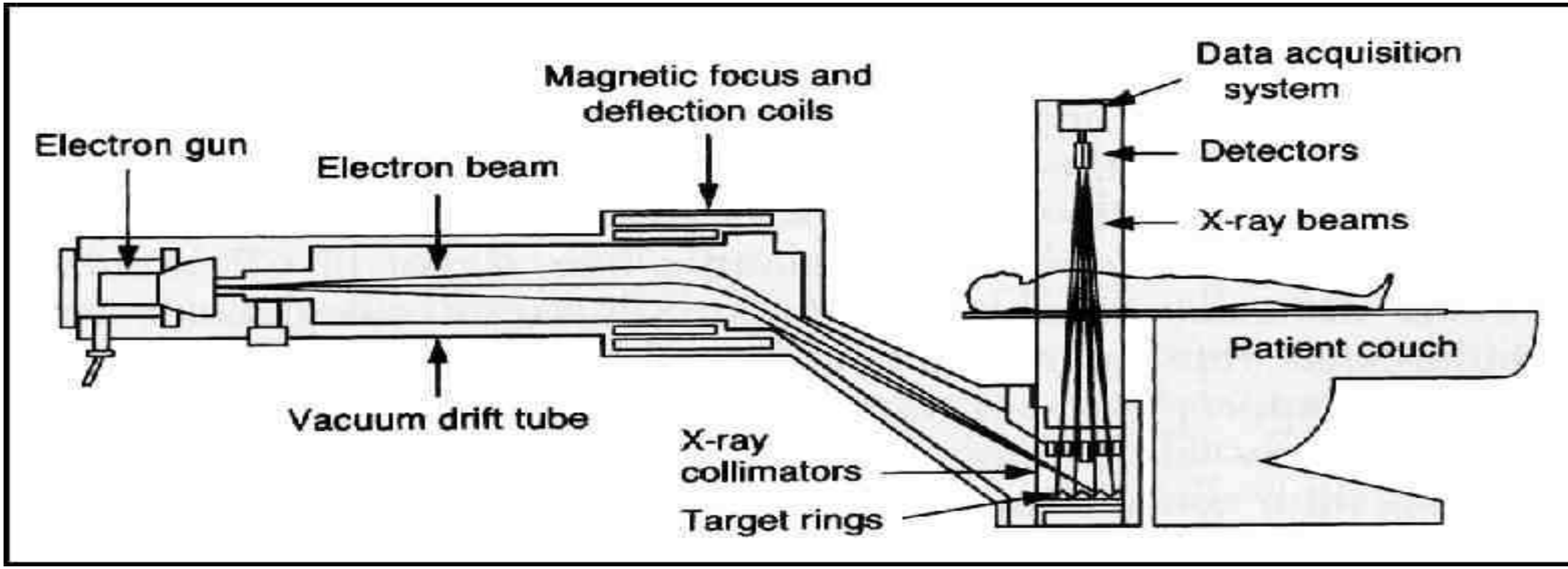
$$3^{rd}\ \text{generation}: \ln(g_1 I_0 / g_2 I_t) = \mu t$$

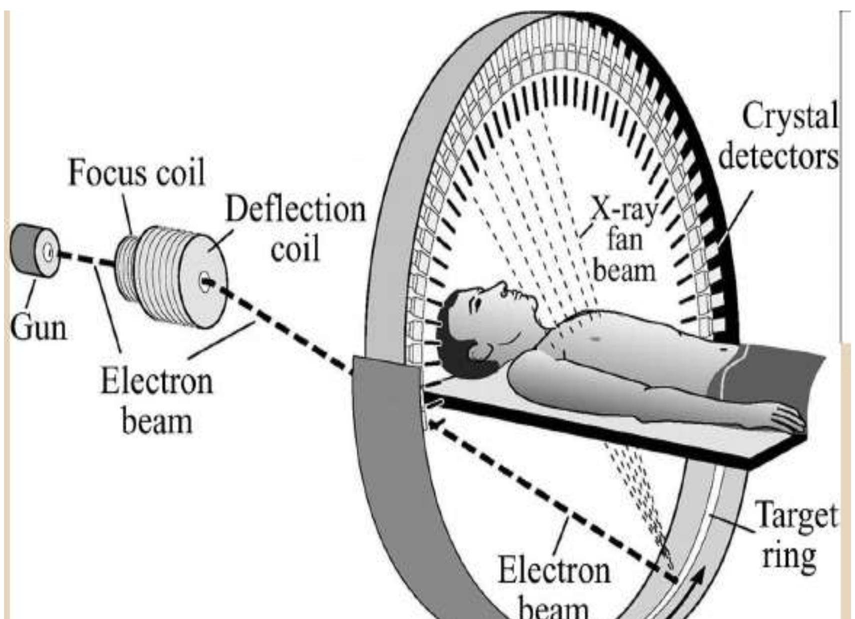
$$4^{th}\ \text{generation}: \ln(g I_0 / g I_t) = \mu t$$

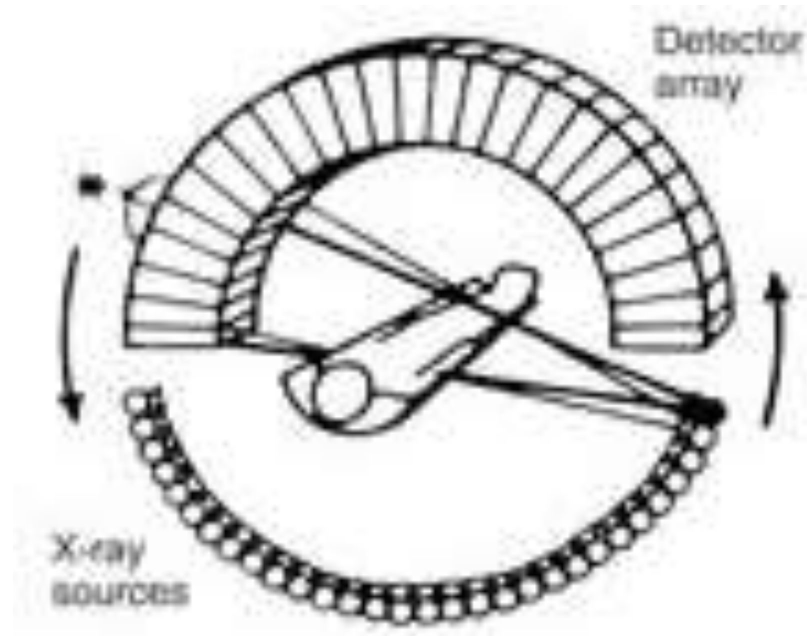
5th generation **(1984)**: stationary/stationary:

Design: No conventional X-ray tube, is a large arc of tungsten encircles patient and lies directly opposite to the detector ring

- No mechanical scanning motion **(**X-ray detector and tube anode are stationary**)**
- Anode, is a very large four semicircular tungsten target ring that forms an arc around the patient scan circle **(**spanning **210** degrees about the patient)
- Source of X-rays is moved around the same path as a fourth generation CT scanner by steering an electron beam around the X-ray anode
- Multiple detectors of two detector banks, eight slices of the patient may be imaged without moving patient (fixed inside of the gantry)
- Terms millisecond CT, ultrafast CT and electron beam CT have also been used, although the latter can be confusing since the term suggests that the patient is exposed to an electron beam
- Electron beam steered around the patient to strike the annular tungsten target
- Very fast scanner, data collection for 1slice is **50-100** ms
- Requires no mechanical motion to acquire data
- Sweeps an intense electron beam across a large, stationary anode target which surrounds the patient
- X -rays produced = high -energy electron beam
- X-rays are emitted from the point where electrons strike target
- X-rays transmitted through object are measured by a stationary array of detectors (scanner gantry)
- Stationary/stationary geometry
- Use: for cardiac tomographic imaging **"cine CT"**
- Cine CT systems, have higher noise level and lower spatial resolution but are ideal for some clinical application
- Cardiac imaging with and without the use of contrast agents, lung imaging, and paediatric studies


Data acquisition
system
Magnetic focus and
deflection coils
Electron gun
Electron beam
Detectors
X-ray beams
Patient couch
Vacuum drift tube
X-ray
collimators
Target rings


Crystal
detectors
Focus coil
Deflection
coil
X-ray
fan
beam
Gun
Electron
beam
Target
ring
Electron
beam


Detector
array
X-ray
sources

6th generation **(1989)**: Spiral/Helical CT

Design: Single X-ray tube (very high power) rotates as patient table is moved smoothly and single-row detector

- Simultaneous source rotation, table translation and data acquisition
- Produces one continuous volume set of data for entire region
- Data for multiple slices from patient acquired at 1sec/slice
- Never-stop and one-direction rotating X-ray tube, detectors
- Slip ring replaced with the X-ray tube voltage cables enable continual tube rotation.
- Allowed 3D image acquisition within a single breath hold
- By avoiding the time required to translate the patient table, the total scan time required to image the patient can be much shorter
- Allows the use of less contrast agent and increases patient throughput
- In some instances the entire scan be done within a single breath-hold of the patient

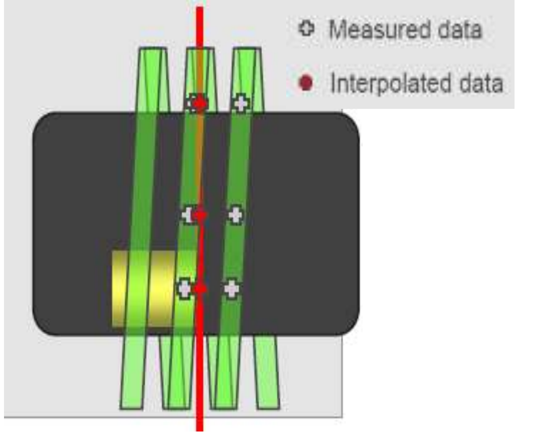
Advantages of Spiral:

- **Speed:** patient movement continuous, shorter exam time and entire object/patient
- **Improved detections**
- **Improved contrast:** image a region in a short period, contrast can be timed
- **Improved reconstruction & manipulation:** volume of data collected, transverse data can be reconstructed in any plane-strip away skin, muscles, etc….

Helical: Continuous gantry rotation and table movement

- Helical CT scanning enables scanners with a single detector to acquire images through an organ or area of interest in less than 20 seconds compared to aver 2 minutes for non-helical scanners.
- This rapidity of acquisition has several advantages.
- Possibility of breath hold acquisitions;
- Intense and homogeneous vascular enhancement with the use of a reduced volume of contrast material.
- Possibility of multi-phase acquisitions through an organ or area of interest during different phases of contrast distribution.
- The liver can thus be imaged before administration of contrast and during the hepatic arterial, portal venous and equilibrium phases of IV contrast distribution, hence improving lesion detection and characterization.
- Another advantage of helical acquisitions is that selection of the reconstruction increment (distance between the center of two contiguous reconstructed images) does not interfere with the acquisition time or patient exposure.
- Overlapping images can thus be reconstructed to improve Z-axis resolution and improve detection of smaller lesions while limiting patient exposure.

Helical Interpolation:



➢ Interpolated helical scan reduces artefacts due to changing structure in z-axis.

➢ For any set reconstruction position, only one scan projection will be at that point.

➢ Interpolation averages data either side of the reconstruction position to estimate projection data at that point.

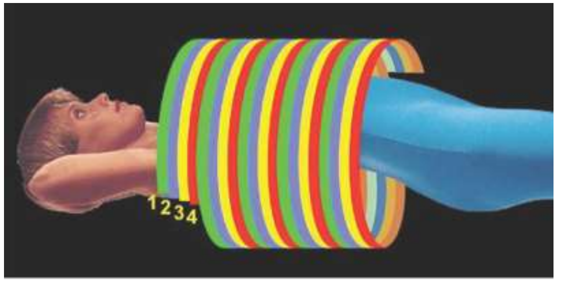
Helical pitch:

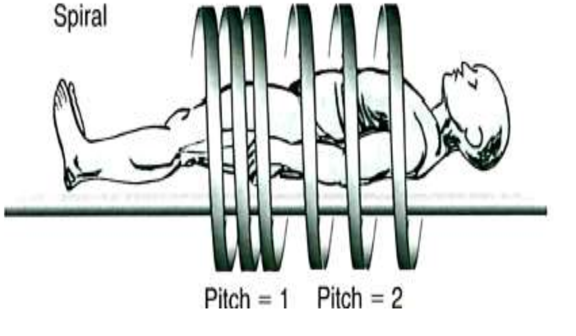
Speed of table movement through gantry defines spacing of helices

$$\text{Pitch} = \frac{\text{couch movement per rotation}}{\text{beam collimation}}$$

- ○ pitch = 1 - coils of the helix are in contact
- ○ pitch < 1 - coils of the helix overlap
- ○ pitch > 1 - coils of the helix are separated

Pitch = 0





Advantages of helical mode:

➢ Pitches greater than 1 possible

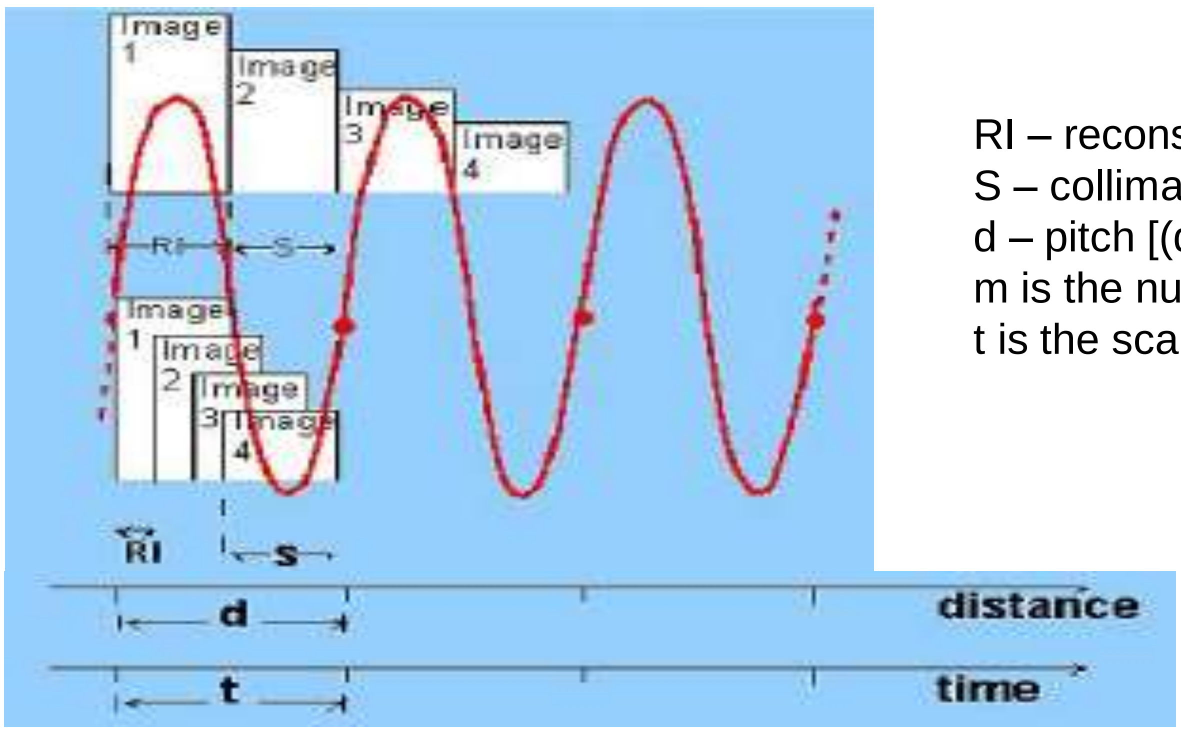
➢ Reduced patient movement

➢ More information more radiation, less information less radiation

➢ Post processing is not possible

Effects of increasing pitch:

- ❑ aster scan time for a specific volume body.
- ❑ Dose is reduced because radiation is less concentrated
- ❑ Image resolution might be reduced
- ❖ when the pitch is increased, table appears to move faster along the patient's body



RI – reconstruction increment (mm)
S – collimation width (mm)
d – pitch [(d/(m x s)]
m is the number of slices
t is the scan time (per 360°) in sec

7th generation **(1998)**: multiple detector array

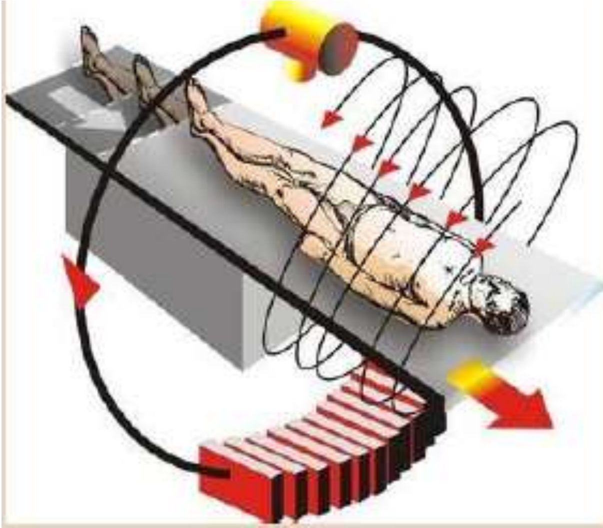
Design: Single X-ray tube (cone beam-widened), multiple parallel rows of detectors array and rotating movement

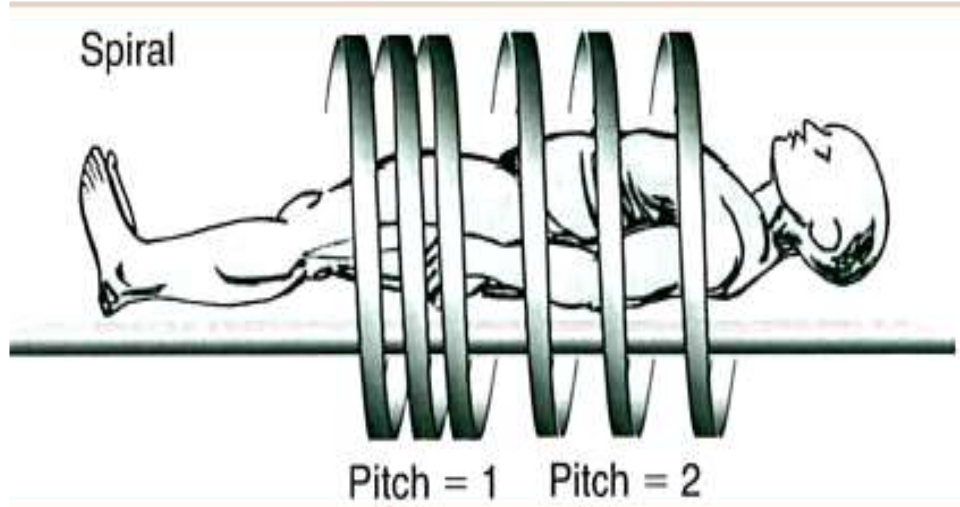
- The collimator spacing is wider and more of the x-rays that are produced by the tube are used in producing image data
- Opening up the collimator in a single array scanner increases slice thickness, reducing spatial resolution in the slice thickness dimension
- With multiple detector array scanners, slice thickness is determined by detector size, not by the collimator

Advantage: reducing scan time/ increase z-resolution

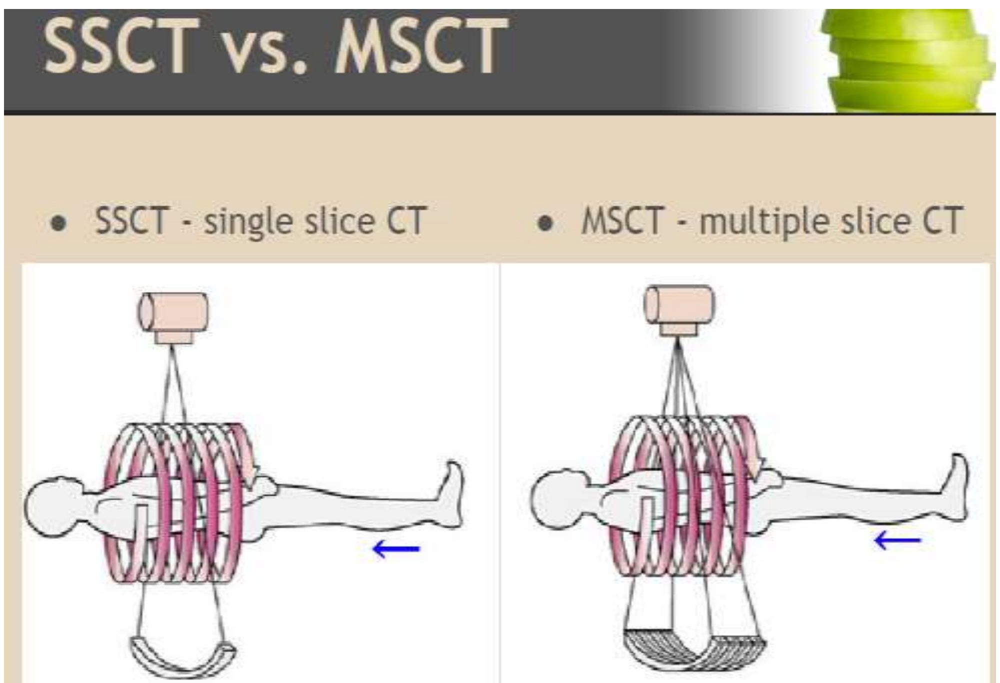
Disadvantage: less scatter rejection compared to single slice, very expensive

- When using multiple detector arrays, the collimator spacing is wider and more of the x-rays that are produced by the tube are used in producing image data.
- Opening up the collimator in a single array scanner increases slice thickness, reducing spatial resolution in the slice thickness dimension.
- With multiple detector array scanners, slice thickness is determined by detector size, not by the collimator.

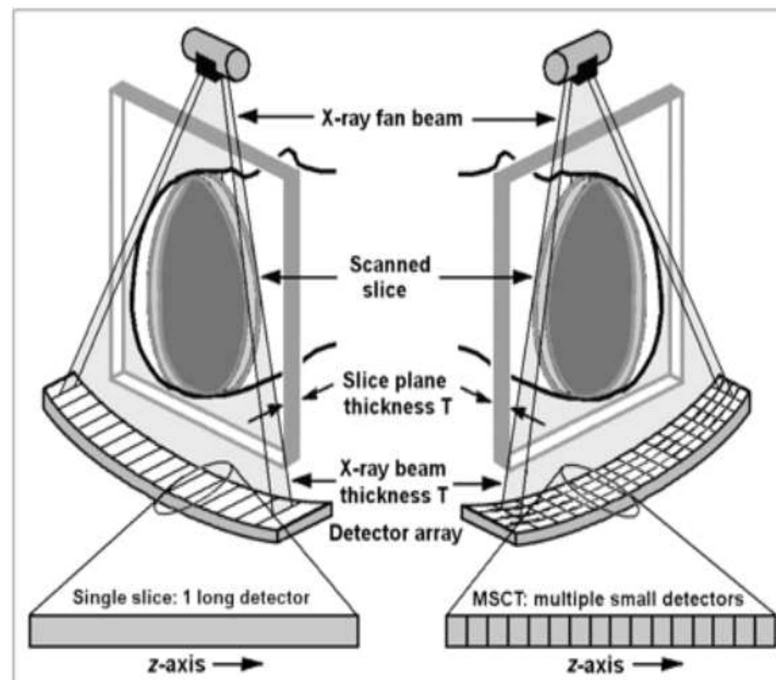




Spiral
Pitch = 1
Pitch = 2



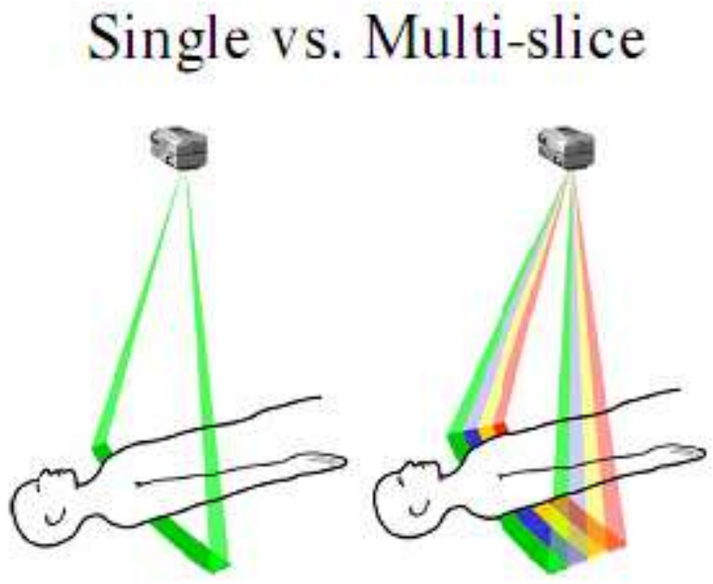
SSCT vs. MSCT
SSCT - single slice CT
MSCT - multiple slice CT



X-ray fan beam
Scanned slice
Slice plane thickness T
X-ray beam thickness T
Detector array
Single slice: 1 long detector
MSCT: multiple small detectors
z-axis
z-axis

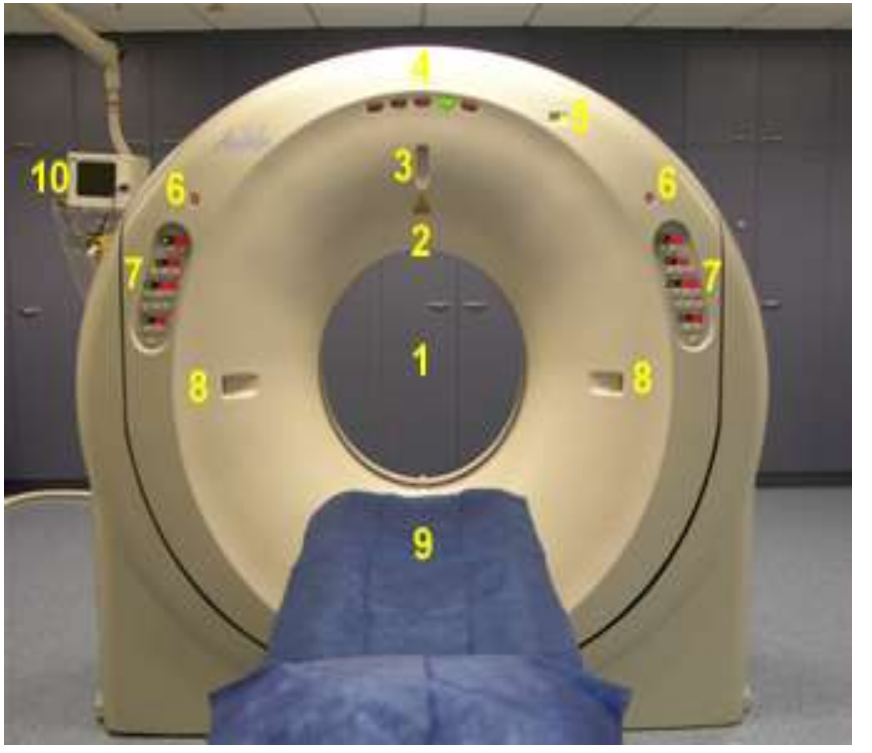
Multiscanning:

- Relates to the technique of double or triple rotation of the tube and detectors around the same axial plane
- Provides double of triple the volume per slice, upon which the final image can be derived
- In practice each rotation produces its own bank of raw data,
- Hence motion which may occur during one rotation can be averaged out from data of the remaining two rotations
- Multiscanning therefore reduces motion artifacts and consequently improves image quality



Modern CT scanner:

1) Gantry aperture (720 mm diameter)
2) Microphone
3) Sagittal laser alignment light
4) Patient guide lights
5) X-ray exposure indicator light
6) Emergency stop buttons
7) Gantry control panels
8) External laser alignment lights
9) Patient couch
10) ECG gating monitor



Gantry: **Internal structure**

- ✓ X-ray Tube
- ✓ Filters, collimator, reference detector and detectors
- ✓ Internal projector, Rotation control unit
- ✓ Data acquisition system (DAS), Slip rings
- ✓ Analog – to – Digital converter (ADC)

Detectors:

- Function as image receptors for remnant/scatter radiation.
  - Its converts the measurement into an electrical signal proportional to the radiation intensity.
- Two types of detector systems are suitable for CT machines:
  - A single multi-chamber inert gas (xenon)/Ionization detector.
  - Multiple scintillation/solid state detectors with photomultipliers/photodiodes detectors.
- Xenon gas detectors: Pressurized xenon gas capable of higher count rates, but lower detection efficiency.
- Modern CT scanners use solid state detectors that have very high efficiency at the low energy of x-rays produced by CT scanners.
- Solid state detectors are made of a variety of materials that create a semiconductor junction similar to a transistor.
- Ultrafast ceramic detectors use rare earth elements such as silicon, germanium, cadmium, yttrium or gadolinium, which create a semiconducting p-n junction.
- Ceramic solid-detectors are very fast, can be extremely stable, and are produced to form an array of very small, efficient detectors that can cover a large area.
- Modern ceramic scintillators : Coupled to photodiodes, offer best all round performance (used to record photon activity).
- Materials: Scintillation crystals such as Cadmium tungstate ($CdWO_4$), Cesium iodide (CsI), Thallium doped Sodium iodide (NaI(Tl)), Ceramic materials containing rare-earth oxides (yttrium or gadolinium), Bismuth germanium oxide (BGO).

Reason for choosing this materials:

- Density and effective atomic number are higher.
- Better x-ray absorption efficiency.

Early detectors scintillators (Ex: NaCl):

- ✓ It is low maximum count rate leads to long scan times.

Gas proportional detectors:

➢ The xenon gas used to fill the chambers has a high atomic number (nonradioactive) which increases photoelectric absorption in the detector.

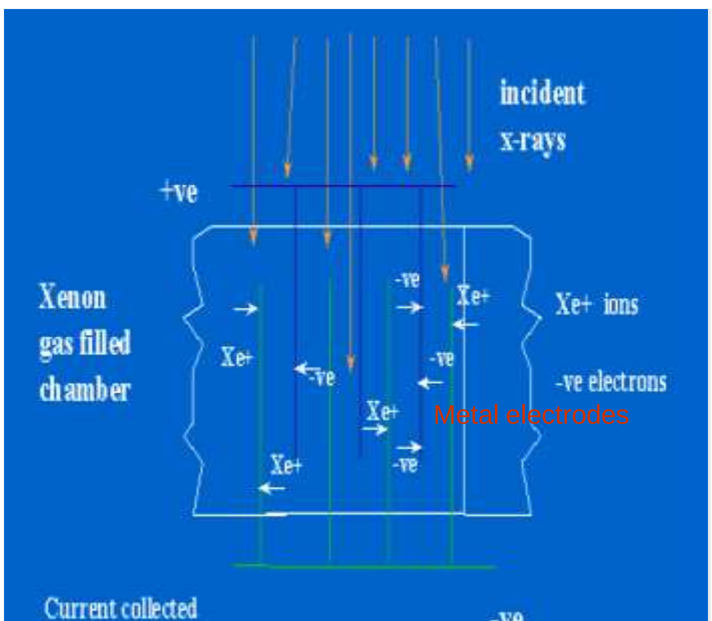
➢ Xenon gas, which is less dependent on stable high voltage supplies, is inherently uniform, and provides in-built collimation as an added bonus.

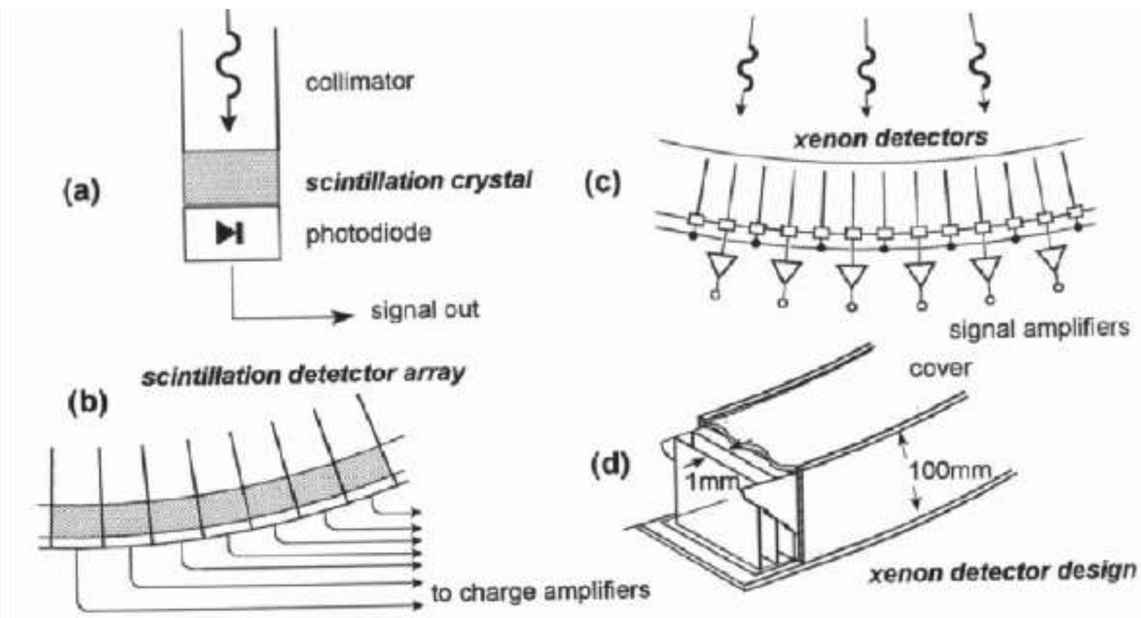
➢ Absorption is further increased by keeping the gas under high pressure and increasing the length of the chambers.

➢ Under these circumstances the sensitivity can be about 50% of the scintillation detector.

➢ The complete detector array is subdivided, using two metal electrode plates, with a voltage applied across the two electrodes.

➢ The electrode plates forming the chambers also act as collimators so added collimation is minimal between patient and detector.

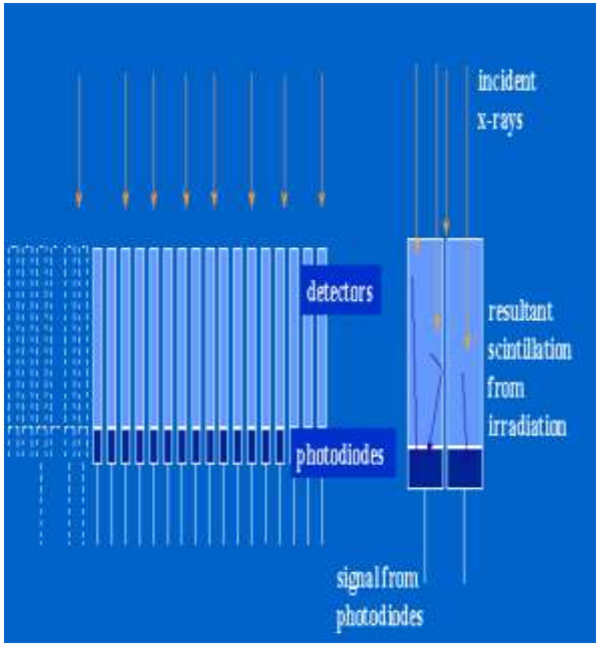


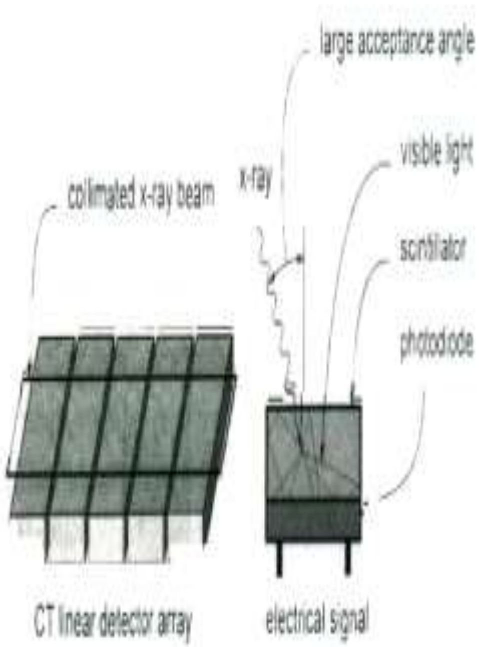


- Each one shares the same gas volume minimizing variation in sensitivity between the chambers.
- As x-rays interact with the xenon atoms and cause ionization (positive atoms and negative electrons), the electric field (volts per centimeter) between the plates causes the ions to move to the electrodes, where the electronic charge is collected.
- The electronic signal is amplified and then digitized, and its numerical value is directly proportional to the x-ray intensity striking the detector.
- The signal strength is not influenced by small supply voltage variations, unlike the crystal/photomultiplier/photodiode systems.
- Xenon detector technology has been surpassed by solid-state detectors, and its use is now relegated to inexpensive CT scanners.
- Each detector anode feeds a dedicated amplifier which is switched to a common A/D converter.
- The detector electrodes are aligned with the focal spot of the x-ray tube.
- This is the acceptance angle of the detector decided by the detector collimation and detector aperture size.
- Xenon detectors can be used only for 3rd generation CT systems.
- Xenon detectors cannot be used for the 4th generation scanner, because those detectors have to record x-rays as the source moves over a very wide angle.
- The fixed detector design of the fourth generation machine has two geometrical centers, that of the gantry–patient and that of the x-ray focal spot.
- This geometric misalignment between x-ray beam and the center of the detector array gives misaligned beam angles at the detector surface so only shallow depth detectors (photodiodes) can be used to accommodate this differing geometry and the packing density of the detectors (influencing resolution) is restricted.
- For this reason the fixed detector design has become unpopular in spite of its fast scan times.

Solid-State Detectors:

- A solid-state CT detector is composed of a scintillator coupled tightly to a photodetector.
- Cesium iodide (CsI) super seded sodium iodide (NaI), as a detector since it gives light photons in the visible range (sodium iodide gives ultraviolet) which can be detected by a simple photodiode which is packed into a very small volume array.
- The scintillator emits visible light when it is struck by x-rays, just as in an x-ray intensifying screen.
- The light emitted by the scintillator reaches the photodetector, typically a photodiode, which is an electronic device that converts light intensity into an electrical signal proportional to the light intensity.
- The photomultiplier tubes could not provide the packing density that was necessary with fan-beam designs so photodiode detectors were substituted.
- This scincillator-photodiode design of solid-state CT detectors is very similar in concept to many digital radiographic x-ray detector systems; however, the performance requirements of CT are slightly different.
- The detector size in CT is measured in millimeters.





➢ The scintillator used in solid-state CT detectors varies among manufacturers, with CdWO4, yttrium and gadolinium ceramics, and other materials being used.

✓ Because the density and effective atomic number of scintillators are substantially higher than those of pressurized xenon gas, solid-state detectors typically have better x-ray absorption efficiency.

➢ However, to reduce crosstalk between adjacent detector elements, a small gap between detector elements is necessary, and this reduces the geometric efficiency somewhat.

Major problems with scintillation detectors are:

✓ Relatively long afterglow following the detection of an x-ray photon.

✓ Stable output signals depend on a very stable high-voltage supply.

✓ Multi-detector uniformity of response is difficult to maintain.

Capabilities of Single Row Detector CT (SDCT):

➢ **Large tissue volumes scanned in short times.**

➢ **Inter-scan delay eliminated.**

➢ **Arbitrary section position within scanned volume permits over-sampling without increased dose.**

➢ **Z axis resolution improved by over-sampling.**

Limitations of Single Row Detector:

➢ **Large volume scan in short duration is limited.**

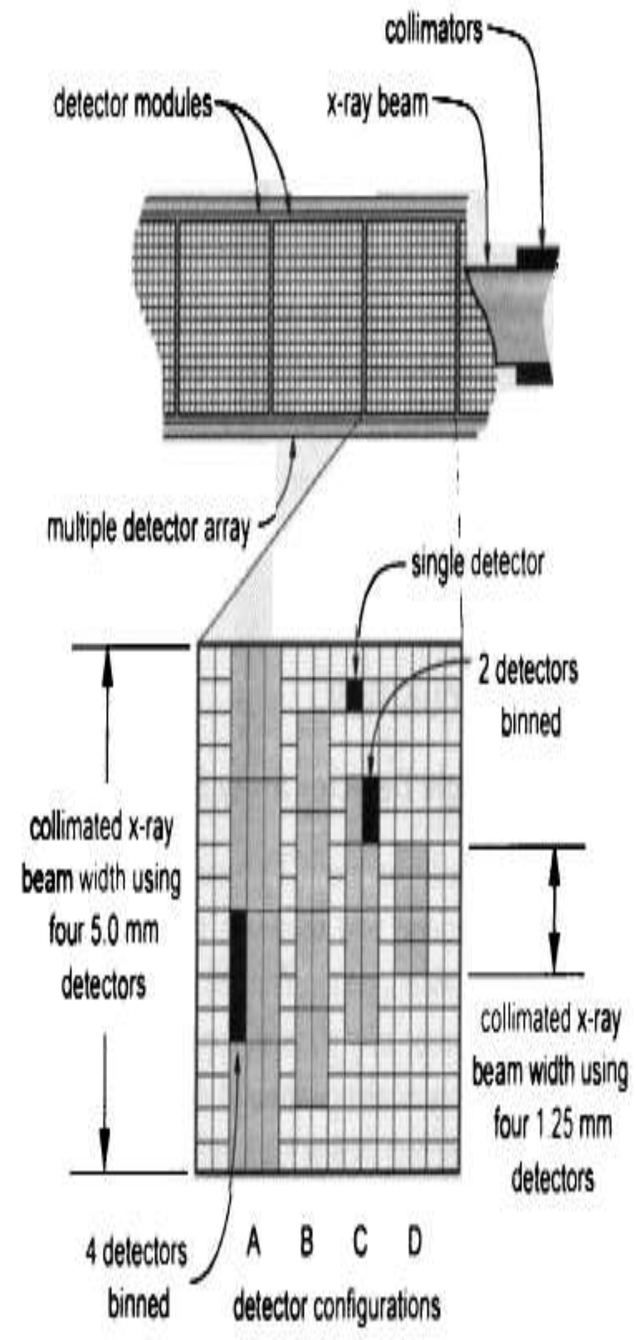
➢ **Near isotropic resolution only over small volume.**

➢ **Poor utilization of x-ray tube.**

➢ **Multiple row detector CT (MDCT) offers substantial improvement in volume coverage, scan speed with efficient use of x -ray tube.**

Multiple Detector Arrays:

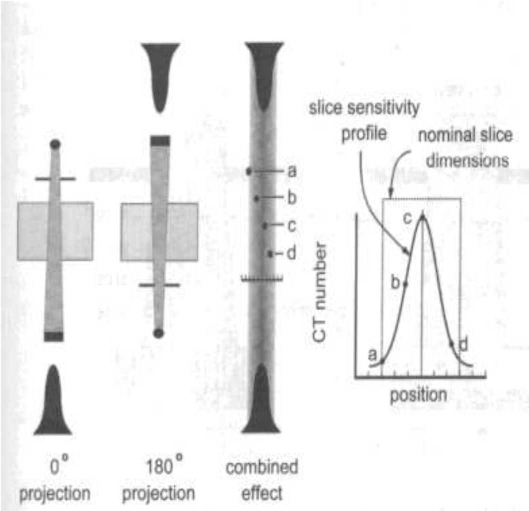
- It is a set of several linear detector arrays, tightly abutted.
- The multiple detector array is an assembly of multiple solid-state detector array modules.
- With a traditional single detector array CT system, the detectors are quite wide and the adjustable collimator determines slice thickness (1 and 13 mm).
- In these systems, the spacing between the collimator blades is adjusted by small motors under computer control.
- Slice width (adjustable) is determined by the detectors (adjustable) but it is not feasible, however, to physically change the width of the detector arrays per sec.
- Therefore, with multislice systems, the slice width is determined by grouping one or more detector units together.
- To combine the signal from several detectors, the detectors are essentially wired together using computer-controlled switches.
- Multiple detector array scanners make use of third-generation geometry.



Details of acquisition:

Slice Thickness: Single Detector Array Scanners

- The slice thickness in single detector array CT system is determined by the physical collimation of the incident x-ray beam with two lead jaws.
  - ✓ As the gap between the two lead jaws widens, the slice thickness increases.
  - ✓ The detector are quite wide and the spacing between the collimators blades is adjusted by small motors under computer control.



- Larger slice thicknesses yield better contrast resolution with the same x-ray techniques, but the spatial resolution in the slice thickness dimension is reduced.
- Thin slices improve spatial resolution in the thickness dimension and reduce partial volume averaging.
- The contrast of a small, highly attenuating ball bearing is greater if the bearing is in the center of the CT slice, and the contrast decreases as the bearing moves toward the edges of the slice. This effect describes the slice sensitivity profile (SSP).
- Single detector array scanners, the shape of the slice sensitivity profile is a consequence of the finite width of the x-ray focal spot, the penumbra of the collimator, the fact that the image is computed from a number of projection angles encircling the patient, and other minor factors.

Slice sensitivity profile (SSP):

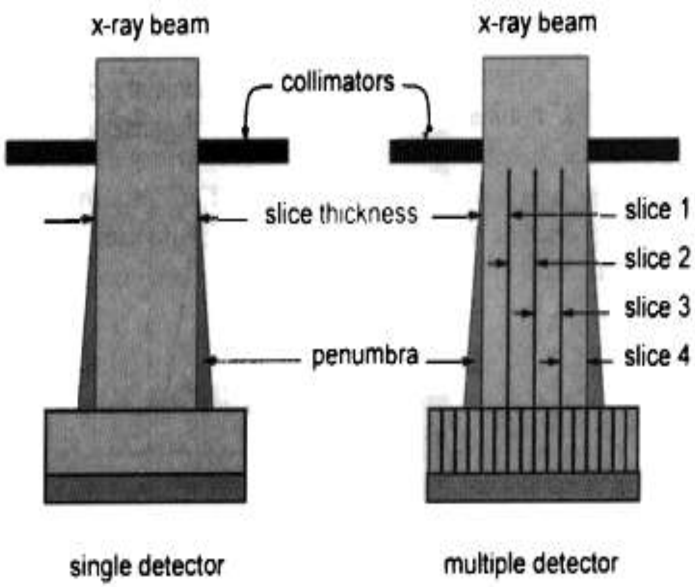
- The sensitivity profile is an important factor of a CT machine since it determines the image quality.
- The steeper the profile slope the less interference from adjacent slices that would cause partial volume effects.
- The perfect sensitivity profile would be rectangular and for a point source of x-rays this could be achieved by simple collimation; however, for a practical system where the focal spot has a finite size geometrical unsharpness causes penumbra effects and tight collimation at the detector entrance is necessary.

Slice dose profile (SDP):

- The slice sensitivity profile and the slice dose profile have different geometries when measured at the axis.
- The larger dose profile contributes to increased surface dose.
- The xenon gas detectors offer built-in collimation so extra detector collimation is unnecessary and there is good agreement between sensitivity and dose profiles.

Slice Thickness: Multiple Detector Array Scanners:

- The multiple detectors array is an assembly of multiple solid-state detector array modules. There are a set of several liner detector arrays with tightly abutted.
- Multiple detectors array, slice width is determined by the detectors not by the collimator.
- To allow the slice width to be adjustable, the detector width must be adjustable. so the electronic signals generated by adjacent detector elements are electronically summed.
- The slice width is determined by grouping one or more detector units together.
- Multiple detector arrays can be used both in conventional axial scanning and in helical scanning prorocols.
- The inner side of the slice is determined by the edge of the detector, but the outer edge is determined either by the collimator penumbra or the outer edge of the detector, depending on collimator adjustment.
- To adjust the collimation (focal spot), collimator blade penumbra falls outside the edge detectors.



Detector Pitch and Collimator Pitch:

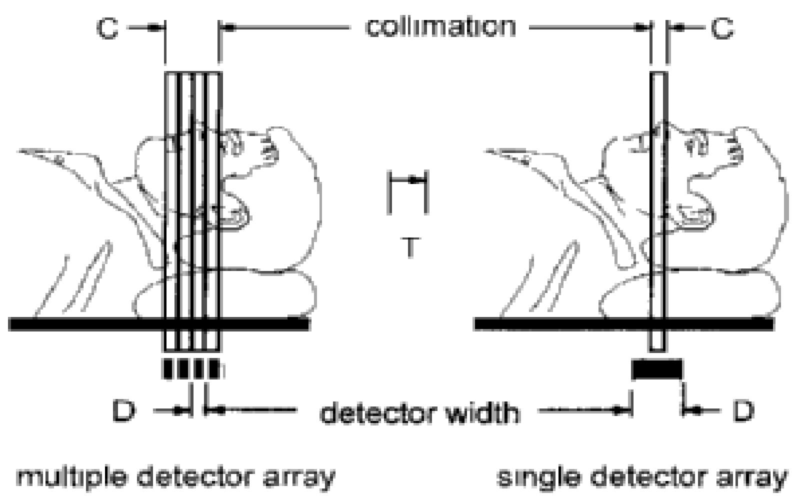
➢ Pitch is a parameter that comes to play when helical scan protocols are used.

✓ In a helical CT scanner with single detector array, the pitch is determined by the collimator and is defined as:

$$\textbf{Collimator pitch} = \frac{\textbf{table movement (mm) per 360 - degree rotation of gantry}}{\textbf{collimator width (mm) at isocenter}}$$

✓ The *detector pitch* is also a useful concept for multiple detector array scanners, and it is defined as:

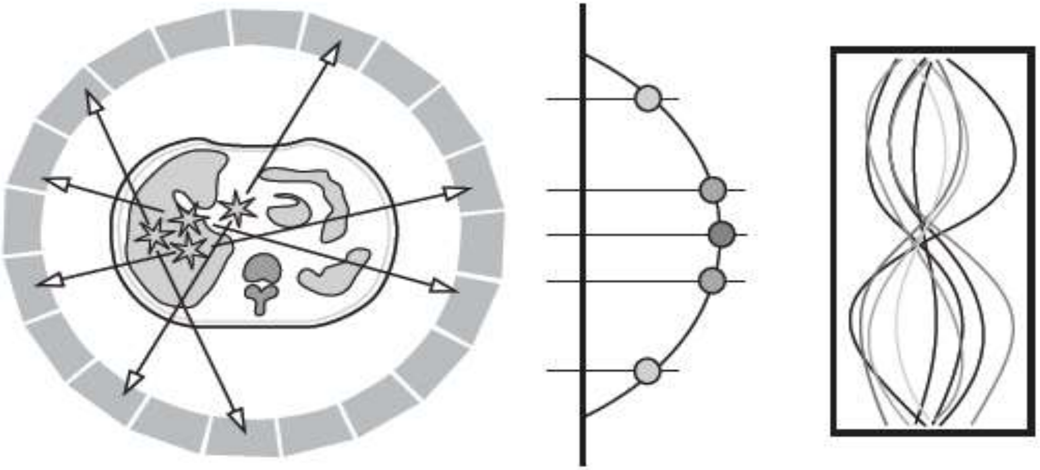
$$\textit{Detector pitch} = \frac{\textit{table movement (mm) per 360 - degree rotation of gantry}}{\textit{detector width (mm)}}$$



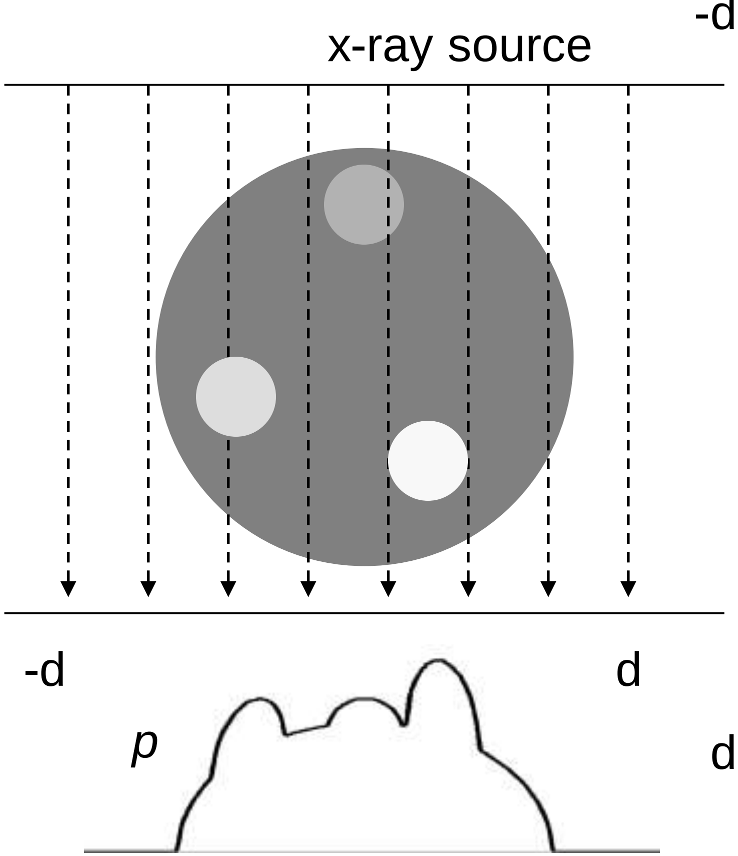
Tomographic reconstruction:

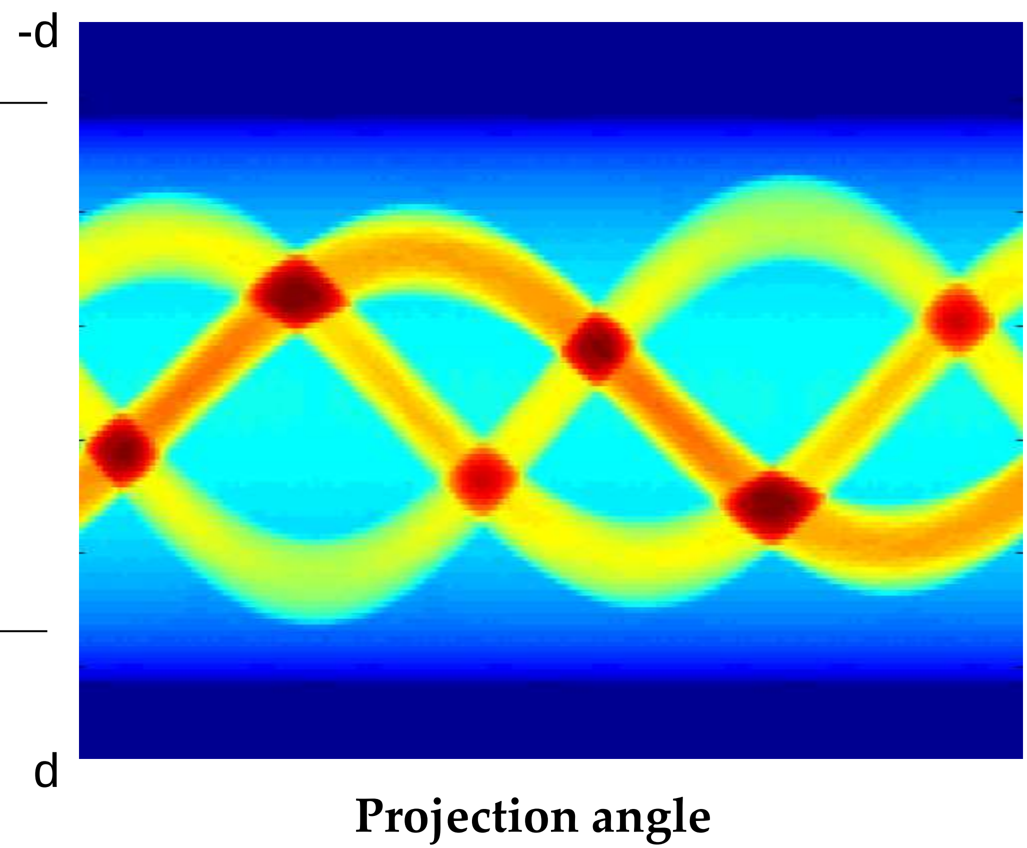
Rays and Views: The Sinogram

- The data acquired for one CT slice can be displayed before reconstruction. This type of display is called a sinogram.
- Sinograms are not used for clinical purposes.
- Coincident events (line-of-response or LOR) for all projection rays are stored into sinograms.
- 2-D slice; a collection represents the entire image.
- The phase and amplitude of the sine wave is unique to the **source location in the tomographic** plane.
- The intensity of the sine wave indicates **source strength.**
- **The sinograms are transferred to the array** processor and used for transaxial image reconstruction using a standard spatial filter.
- Image construction from these sinograms then relies on filtered backprojection or iterative reconstruction.



Projections (p) = ln (I_o/I_t),
where I_o – reference data,
I_t – corresponding to each ray
and
t – thickness





Projection angle

Interpolation (Helical):

➢ Helical CT scanning produces a data set in which the x-ray source has traveled in a helical trajectory around the patient.

✓ Present-day CT reconstruction algorithms assume that the x-ray source has negotiated a circular, not a helical, path around the patient.

✓ To compensate for these differences in the acquisition geometry, before the actual CT reconstruction the helical data set is interpolated into a series of planar image data sets.

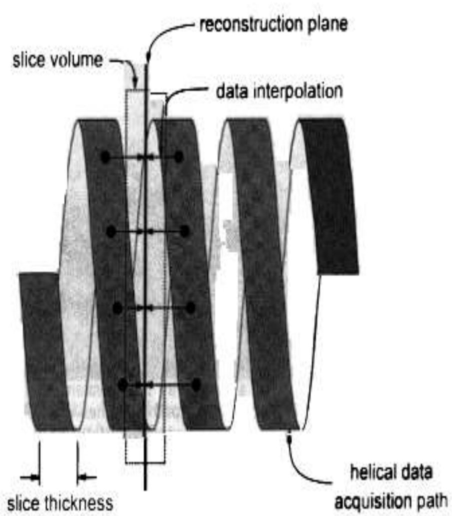
✓ During helical acquisition, the data are acquired in a helical path around the patient. Before reconstruction, the helical data are interpolated to the reconstruction plane of interest.

✓ Interpolation is essentially a weighted average of the data from either side of the reconstruction plane, with slightly different weighting factors used for each projection angle.

➢ Although this interpolation represents an additional step in the computation, it also enables an important feature.

✓ With conventional axial scanning. the standard is to acquire contiguous images, which abut one another along the cranial-caudal axis of the patient.

✓ With helical scanning, however, CT images can be reconstructed at any position along the length of the scan to within (½) (pitch) (slice thickness) of each edge of the scanned volume.



➢ Helical scanning allows the production of additional overlapping images with *no additional dose* to the patient.

✓ The sensitivity of the CT image to objects not centered in the voxel is reduced (as quantified by the slice sensitivity profile), and therefore subtle lesions, which lay between two contiguous images, may be missed.

✓ With helical CT scanning. interleaved reconstruction allows the placement of additional images along the patient, so that the clinical examination is almost uniformly sensitive to subtle abnormalities.

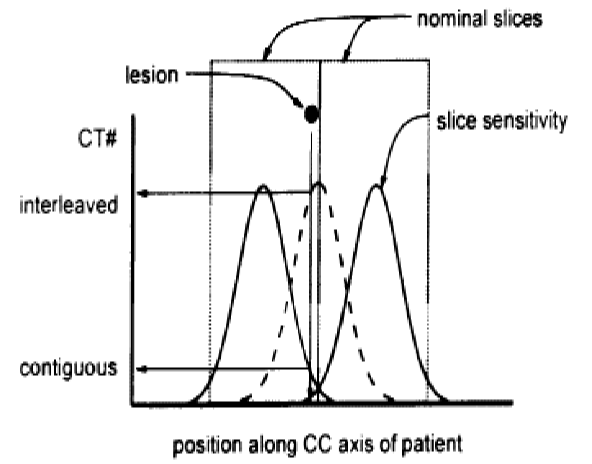
➢ Interleaved reconstruction adds no additional radiation dose to the patient, but additional time is required to reconstruct the images.

✓Although an increase in the image count would increase the interpretation time for traditional side-by-side image presentation, this concern will ameliorate as more CT studies are read by radiologists at computer workstations.

➢ This figure illustrates the value of interleaved reconstruction.

✓ The nominal slice for contiguous CT images is illustrated conceptually as two adjacent rectangles; however, the sensitivity of each CT image is actually given by the slice sensitivity profile *(solid lines).*

✓ A lesion that is positioned approximately between the two CT images *(black circle)* produces low contrast (i.e., a small difference in CT number between the lesion and the background) because it corresponds to low slice sensitivity.



✓ With the use of interleaved reconstruction *(dashed line),* the lesion intersects the slice sensitivity profile at a higher position, producing higher contrast.

➢ It is important not to confuse the ability to reconstruct CT images at short intervals along the helical data set with the axial resolution itseIf.

✓ The slice thickness (governed by collimation with single detector array scanners and by the detector width in multislice scanners) dictates the actual spatial resolution along the long axis of the patient.

Bone Kernels and Soft Tissue Kernels:

➢ The reconstruction filters derived by Lakshminarayanan, Shepp and Logan, and Hamming provide the mathematical basis for CT reconstruction filters.

✓ In clinical CT scanners, the filters have more straightforward names, and terms such as “bone filter” and “soft tissue filter” are common among CT manufacturers.

➢ The term *kernel* is also used.

✓ Bone kernels have less high-frequency roll-off and hence accentuate higher frequencies in the image at the expense of increased noise.

✓ CT images of bones typically have very high contrast (high signal), so the SNR is inherently quite good.

✓ Therefore, these images can afford a slight decrease in SNR ratio in return for sharper detail in the bone regions of the image.

➢ For clinical applications in which high spatial resolution is less important than high contrast resolution—for example, in scanning for metasratic disease in the liver—soft tissue reconstruction filters are used.

✓ These kernels have more rolloff at higher frequencies and therefore produce images with reduced noise but lower spatial resolution.

✓ The resolution of the images is characterized by the modulation transfer function (MTF).

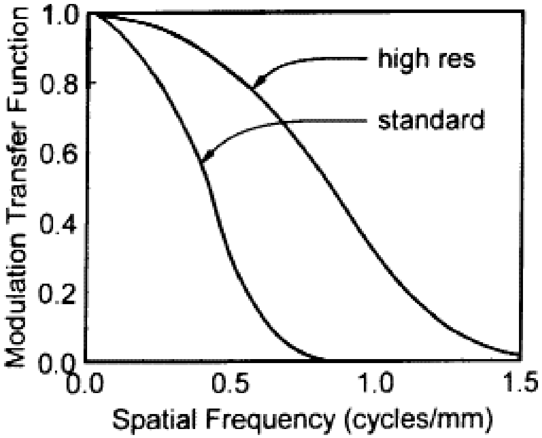
➢ The high-resolution MTF corresponds to use of the bone filter at small field of view (FOV), and the standard resolution corresponds to images produced with the soft tissue filter at larger FOV.

➢ The units for the X-axis are in cycles per millimeter, and the cutoff frequency is approximately 1 .0 cycles/mm.

✓ This cutoff frequency is similar to that of fluoroscopy and it is five to seven times lower than in general projection radiography.

✓ CT manufacturers have adapted the unit cycle/cm

✓ For example, 1.2 cycles/mm = 12 cycles/cm.



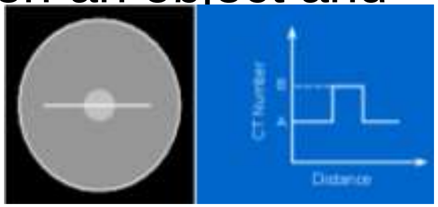
CT fluoroscopy reconstruction:

- CT scan images offer cross-sectional imaging of the target and its surrounding structures, whereas fluoroscopy offers real-time, imaging, tracking any movement of the target, and its reaction to the biopsy needle.
- CT fluoroscopy is similar to conventional CT, with a few added technologic nuances.
- The conventional CT fluoroscopy provides an image sequence over the same region of tissue and therefore the radiation dose from repetitive exposure to that tissue is a concern.
- The benefits are of importance for interventions in the thorax where accuracy is crucial in the diagnosis of small pulmonary, mediastinal, or hilar lesions.
- Real-time observation of the motion of the needle in its trajectory to the intended target maximizes the diagnostic yield and minimizes the need to redirect the needle or perform multiple passes, therefore minimizing the risks of complications.
- Today’s units allow a dose reduction to the patient’s sensitive organs and to the operators hands by automatically switching off the generation of X-rays when they are not needed.

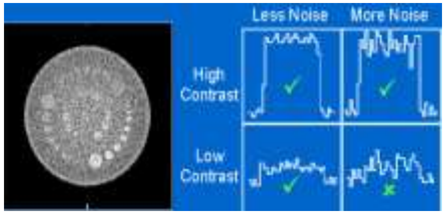
- The latest developments have allowed to obtain a very high rate of images allowing for improved image quality and therefore allowing accurate biopsies of subcentimeter pulmonary nodules.
- Although a whole range of invasive procedures can be performed with the help of CT fluoroscopy, biopsies of pulmonary nodules remain the most common application.
- Other invasive procedures that can be performed by CT fluoroscopy include radiofrequency ablation, laser nucleotomy, vertebroplasty, stereotaxis, and pain therapy.
- CT fluoroscopy is the ideal guiding device for transbronchial biopsies of mediastinal and hilar lesions that are small or difficult to reach.
- It allows for the accurate tracking of the transbronchial device through its trajectory and into its intended target, increasing the diagnostic yield and minimizing the risk of complications by avoiding major structures of the hila and mediastinum.

CT image quality:

- Imaging performance
  - Image noise
    - Variation in CT number in image of a uniform images
    - Results of random processes involved in X-ray interactions and detection
    - Measured using the standard deviation of the image CT numbers
    - Noise is important when looking at low contrast images
  - Contrast
    - Contrast = difference in signal
      = difference in CT number (in Hounsfield Units) between an object and the surrounding tissue $=CT_B-CT_A$



    - When looking at objects which have CT number close to background, noise can mask detail



  - Spatial resolution
  - Z sensitivity
- Patient dose
  - Local. organ and effective doses

Factors affecting image noise:

- Noise is a product of statistical fluctuations in signal at detectors
  - Great signals to detectors – lower noise
- Each detector trying to estimate attenuation
  - Does this by counting X-rays. More X-rays give more accurate attenuation measurement.
- Convolution kernel/reconstruction filter
  - Smooth kernel gives low noise, at expense of spatial resolution

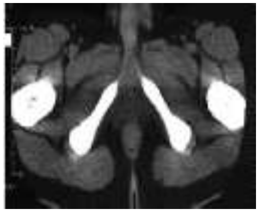
Factors affecting detector signal:

- kV: high kV X-rays more penetrating
- mA: high tube current gives more intense X-ray beam
- Scan time: long scan time, (more X-rays to detectors)
- Slice thickness: wide slice (more X-rays)
- Patient composition: small patients less attenuating

Definition of CT artifacts:

- Systematic discrepancies between the CT numbers in the reconstructed image and the true attenuation coefficients of the object.

  Ex: Non-random or structured, image noise

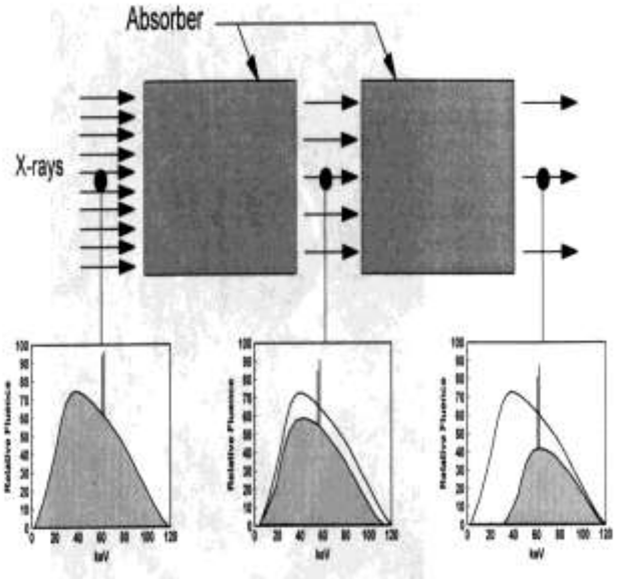


- Common artifacts:

❖ Physics based: Beam hardening and Partial volume effect

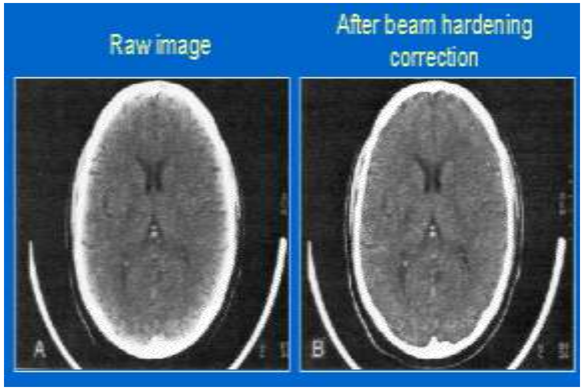
❖ Technique: Bad detector (3^{th} generation scanner)

❖ Patient based: Metal and Patient motion



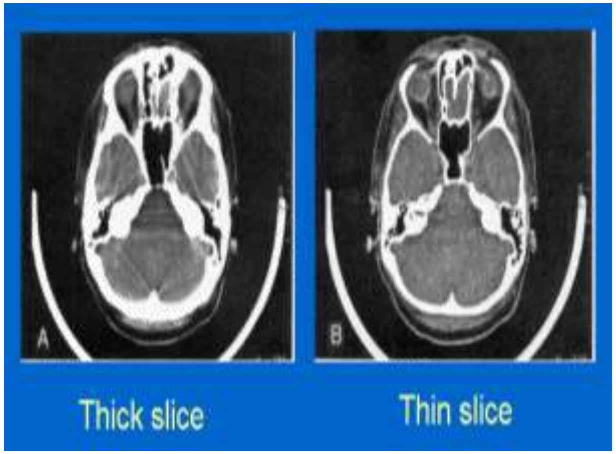
Beam hardening effect:

- Linear attenuation coefficients vary with photon energy.
- After passing through a given thickness of tissue, lower-energy x-rays are attenuated to a greater extent than high-energy x-rays are. i.e., x-ray beam propagates through a thickness of tissue and bone, the shape of the spectrum becomes skewed toward the higher energies.
- Artifacts such as a reduced attenuation toward the center of tissue (cupping) and streaks that connect tissues with strong attenuation.
- Polychromatic spectrums:



Means for suppressing beam hardening effect:

- ❖ Pre-filtering x-rays
- ❖ Avoiding high x-ray absorbing regions if possible
- ❖ Applying appropriate algorithms

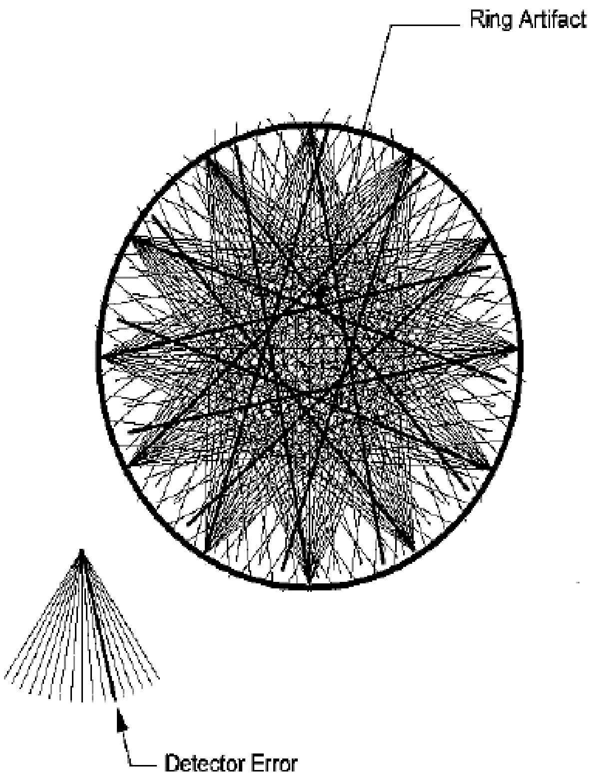


Partial volume effect:

- ➢ Partial volume artifacts are the result of a variety of different tissue types being contained within a single voxel
- ➢ Measured attenuation coefficient are averaged by all components
- ➢ Use thinner slice to reduce

Bad detector (3th generation scanner):

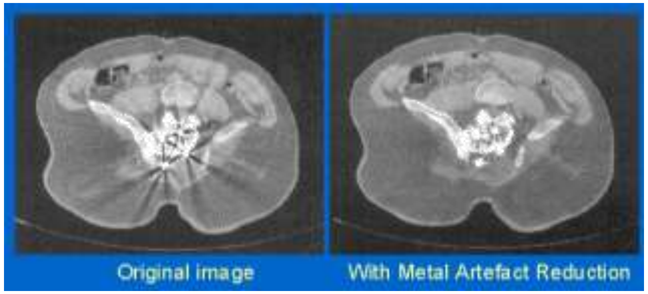
- ➢ Each detector views a separate ring of an anatomy.
- ➢ Any single detector or a bank of detectors malfunctions will produce ring artifact



Metal artifact:

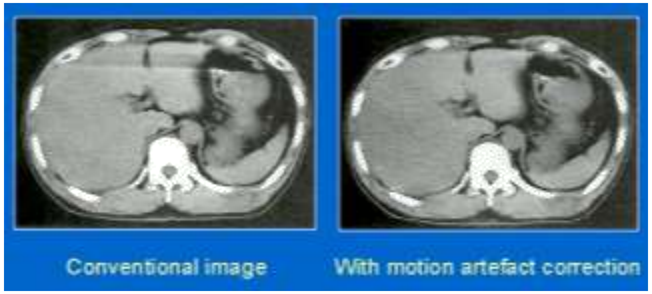
- ➢ Metal materials can cause the streaking artifacts due to block parts of projection data.

Ex: Dental fillings, Prosthetic devices, Surgical clip, Electrodes



Patient/Artifacts motion:

- ➢ Motion artifacts occur when the patient moves during the acquisition.
- ➢ Small motions cause image blurring and large physical displacements during CT image acquisition produce artifacts that appear as double images or image ghosting.



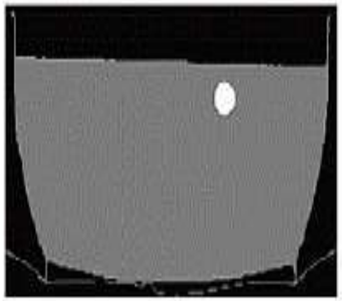
Effect of reducing projections:

- ✓ The number of views (projections)

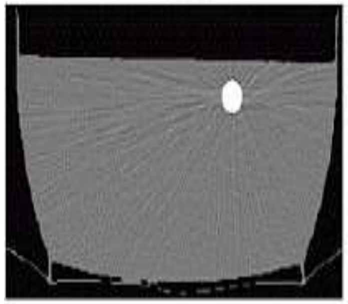
Effect of reducing rays:

- ✓ The numbers of the data point (rays) per projection.

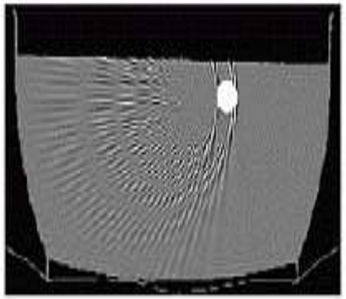
Artifacts simulation



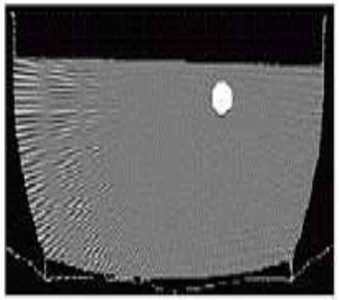
➢ Normal phantom (simulated water with iron rod).



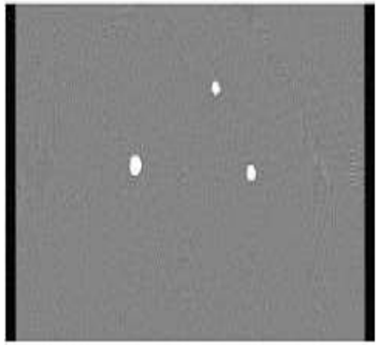
➢ Adding noise to sinogram gives rise to streaks.



➢ Aliasing artifacts when the number of samples is too small (ringing at sharp edges).

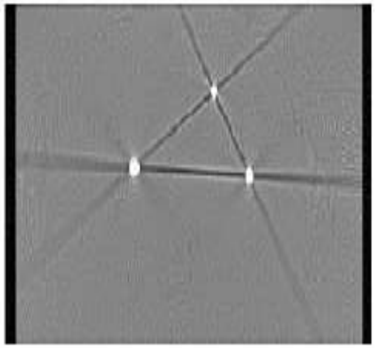


➢ Aliasing artifacts when the number of views is too small.



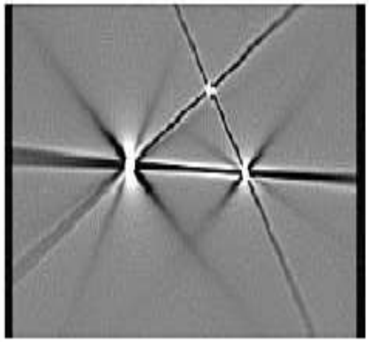
➢ Normal phantom (plexiglas plate with three amalgam fillings).

➢ Beam hardening artifacts



✓ Non-linearities in the polychromatic beam attenuation (high opacities absorb too many low energy photons and the high energy photons won't absorb).

✓ attenuation is under-estimated.



➢ Scatter (attenuation of beam is under-estimated).

✓ the larger the attenuation, the higher the percentage of scatter.

CT artifacts: Summary

- Artifacts originate from a range of sources and can degrade the diagnostic quality of an image
- Some can be partially corrected for in software
- Good scanner design, careful positioning of patient and optimum selection of scan parameters can minimise the artifacts present in an image.

CT vs MRI:

- MRI the profession speculated if CT would become obsolete.
- CT scans are usually cheaper.
- CT scans are typically better at showing bones than MRI, but less effective at showing the soft tissue.
- CT scans take around 5 minutes, MRI's usually take 30 minutes.
- CT scans can be harmful to the patient, while MRI's have no known biohazards.
- Both used for detecting cancer.
- Advantages of CT: Metal is not affected, Claustrophobia, uncooperative patiens, obese patients, fast scan times, trauma, more cost effective

Limitations:

- In order for better clarity, more exposure to radiation is necessary.
- Human error in reading scans.
- Not very good at depicting soft tissue.
- In order to change the image plane you have to move the patient, unlike with MRI.

CT in the future:

- CT technologist has an increased responsibility to understand scanning dynamics.
- This imaging modality will continue to be a highly respected diagnostic tool.