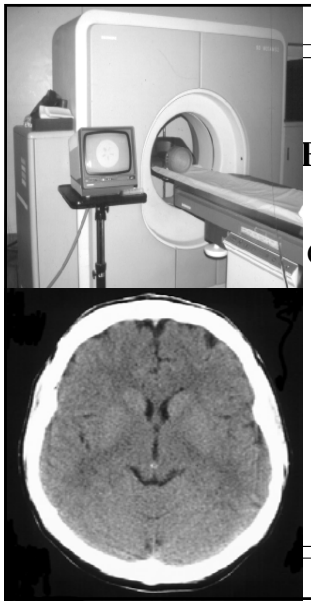

Basic principles of CT scanners and image reconstruction

CT scanning and imaging parameters

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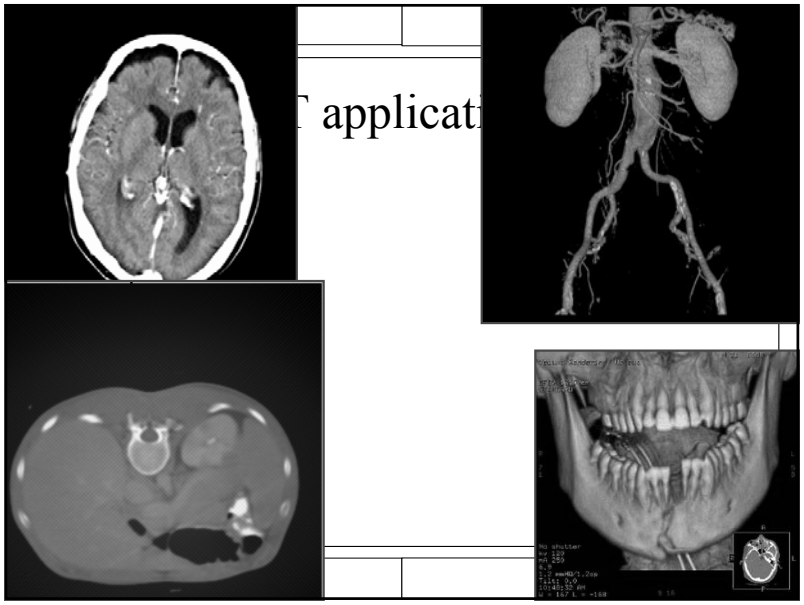
Dept. Medical Eng. and Physics
King's College London

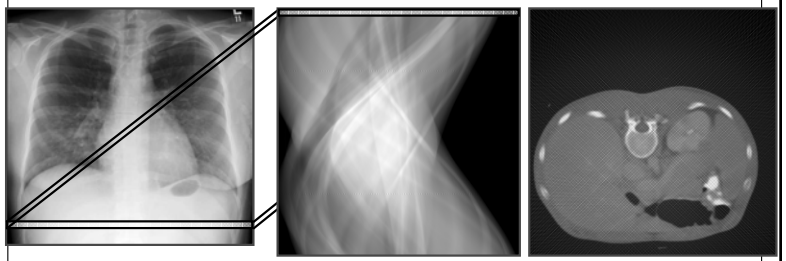


T applicat

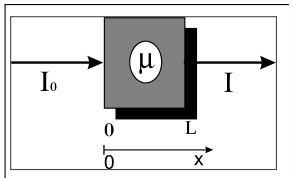
ADVANTAGES:

A number of 2D views (projections) of an object are used to calculate its shape in the 3rd dimension – scan
Differentiate overlying structures

- Overlying structures do not decrease contrast
- Increased contrast
- Digital images with variable window settings



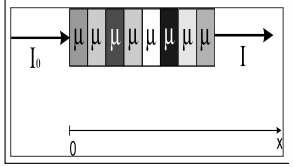
Planar x-ray | Sinogram | Reconstructed image

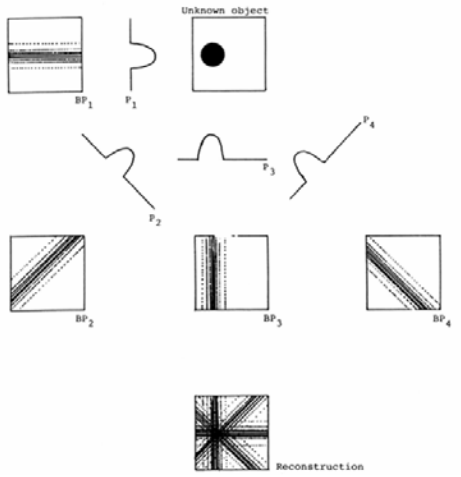


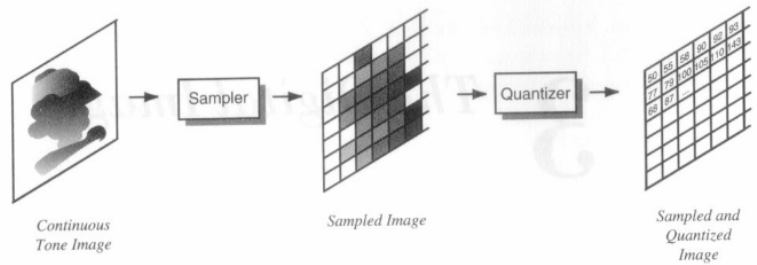
? How to visualise all parts of a "closed" 3D object ?

-voxel: elementary volume

-pixel: element of the image

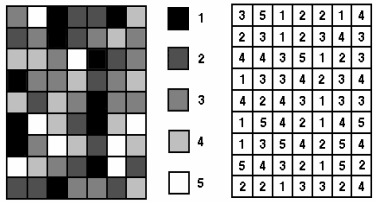




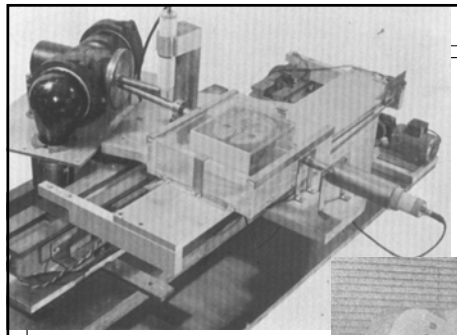


2 types of images :

- invisible digital image "*latent*" (in computer memory)
- visible image in the display (on the computer monitor)



3	5	1	2	2	1	4
2	3	1	2	3	4	3
4	4	3	5	1	2	3
1	3	3	4	2	3	4
4	2	4	3	1	3	3
1	5	4	2	1	4	5
1	3	5	4	2	5	4
5	4	3	2	1	5	2
2	2	1	3	3	2	4

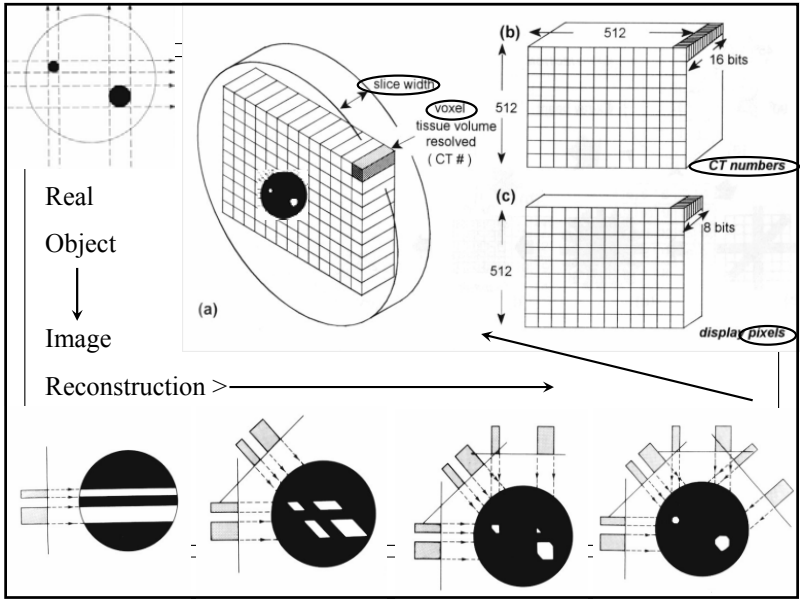


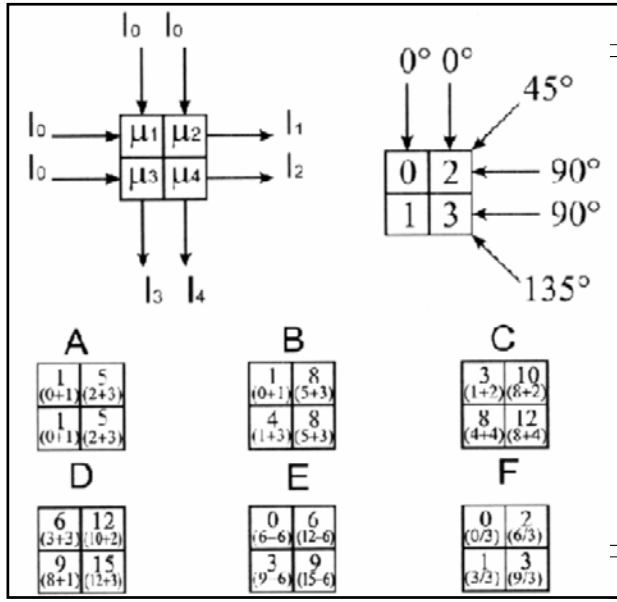
Mathematical theory (Radon,1917)

Applications in astronomy (1950)

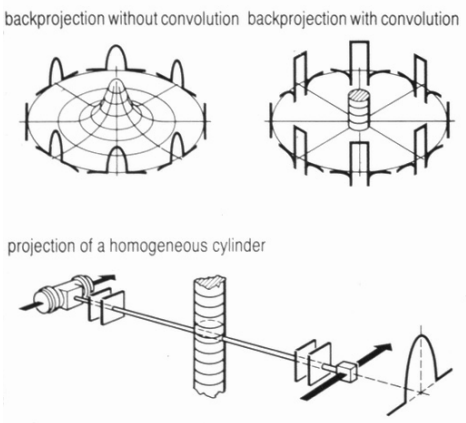
First CT experimental set (1960)

The first CT (EMI MARK1) and Sir Hounsfield - 1973



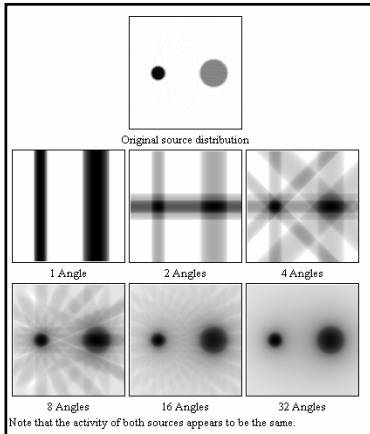


A simple example of reconstruction based on data from 4 angles (projections)



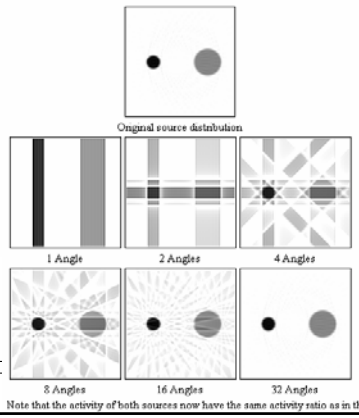
CT image is a product of complex calculation based on backprojection reconstruction algorithm.

The CT image is not a real "shade" like in classical Radiography, but a picture which represents with some probability (>95%) the similarity between the real object and its calculated CT image.

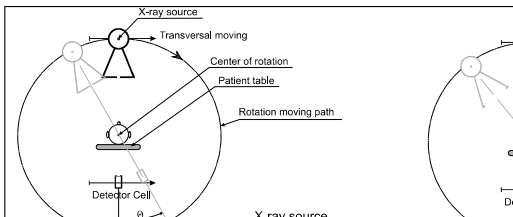


<<< 1. Simple (unfiltered) backprojection

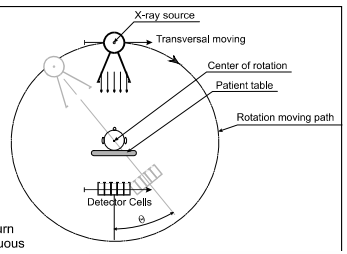
2. Filtered backprojection >>>

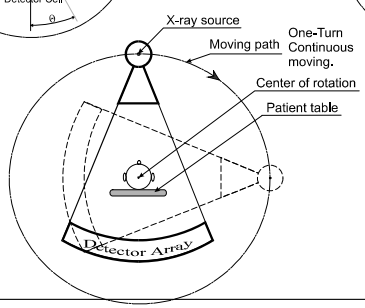


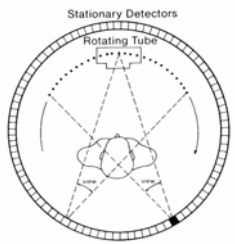
1st Generation



2nd Generation



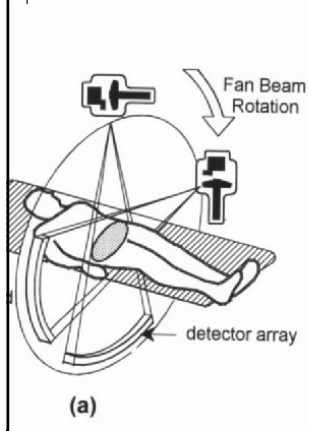


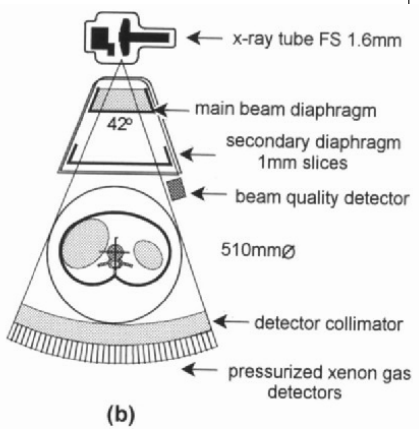


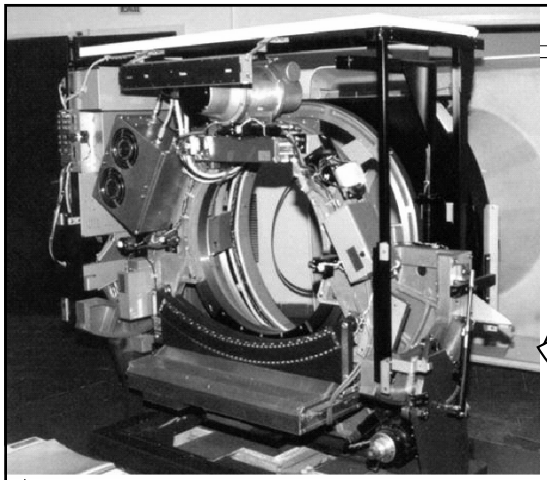
3rd Generation

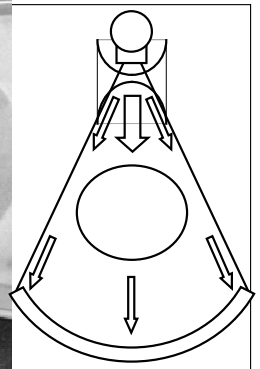
4th Generation;

Typical III generation CT scanner



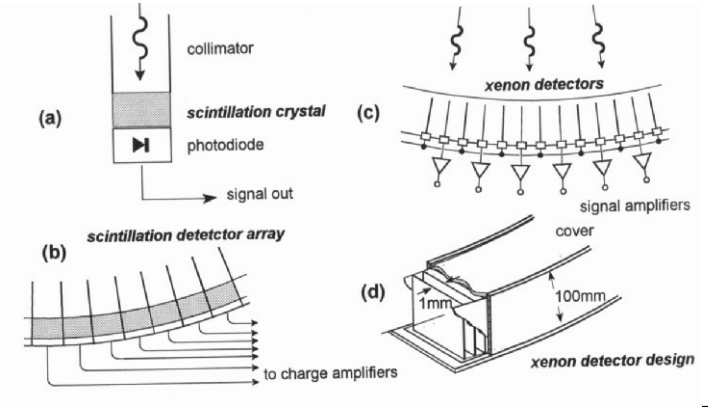




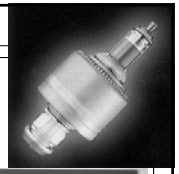


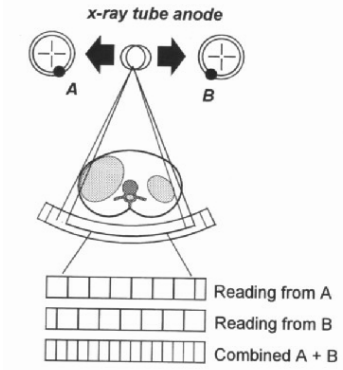
Beam shaping filter ensures more constant signal to all detectors
It also makes Beam hardness (energy) at detectors more constant

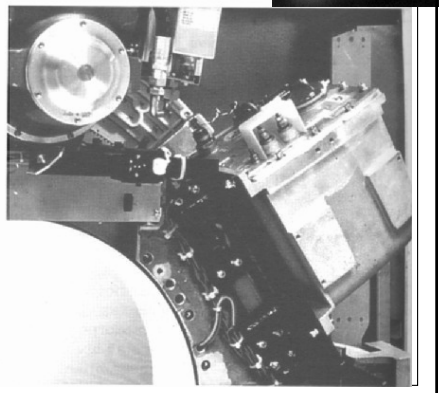
CT detectors development - solid state and gas (Xe) filled
Increased detector efficiency and new (III) generation scanners



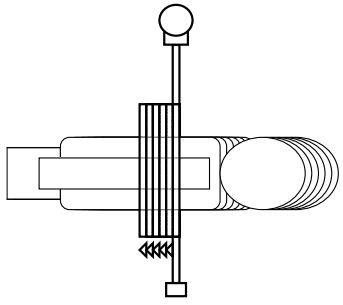
Increase spatial resolution using powerful new
X-ray tubes with flying focal spot technology



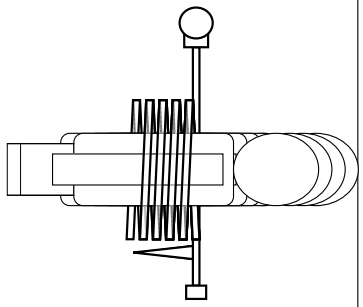




Helical CT - Data Collection



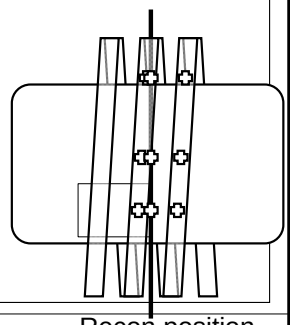
Step and shoot /
sequential mode



Helical / spiral mode

Helical Interpolation

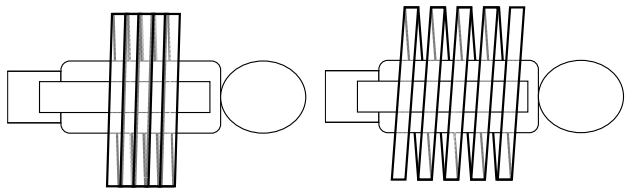
- Interpolated helical scan data reduces artefacts due to changing structure in z-axis
- For any set reconstruction position, only one scan projection will be at that point
- Interpolation averages data either side of the reconstruction position to estimate projection data at that point



Helical Pitch

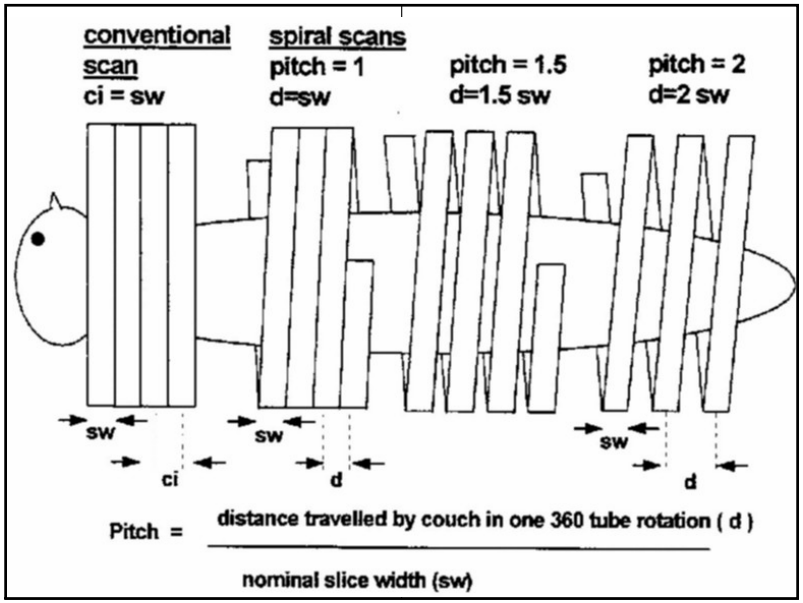
Speed of table movement through gantry defines spacing of helices

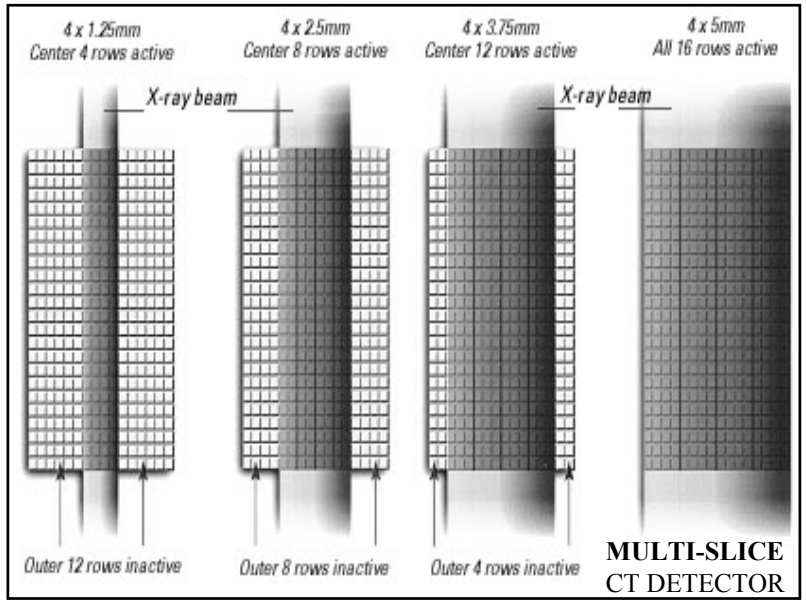
$$\text{Pitch} = \frac{\text{Table travel per rotation}}{\text{x-ray beam width}}$$



Travel = 10 mm/rot
Beam = 10 mm
Pitch = 1

Travel = 20 mm/rot
Beam = 10 mm
Pitch = 2

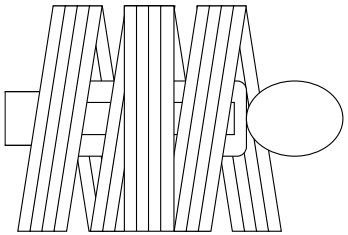


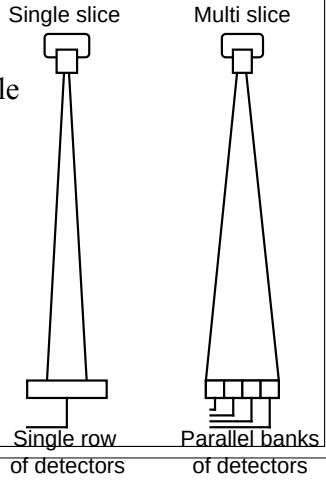


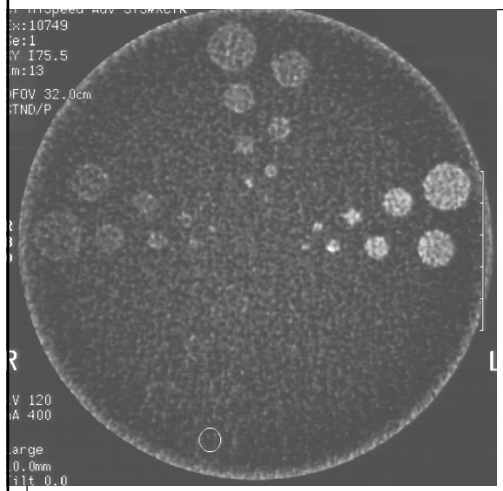
5

Multi-slice CT

- Multi slice detectors
  - introduced 1998
  - allow acquisition of multiple slices in a single rotation



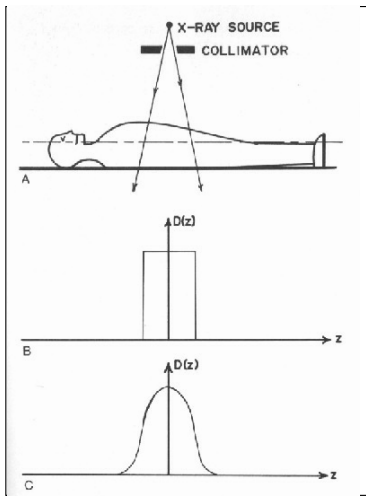




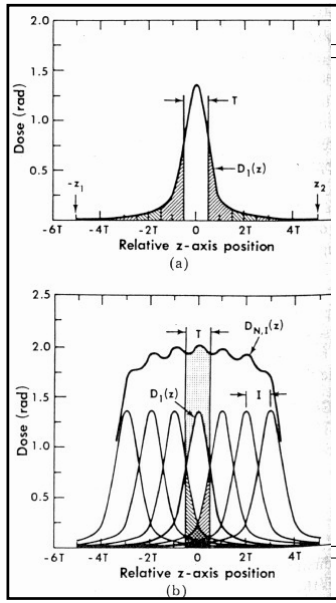
CT contrast resolution is one of the dramatic advantages of the method, compared with classical radiography.

CT discriminates density difference ~ 0,25-0.5 % while normal Radiography discriminates ~ 10%

CT spatial resolution is limited by the image matrix size, detectors and algorithm



The beam width is determined by the tube collimator. Ideally it would be with steep sides, in practice it is spread (bell shape dose distribution). When numerous contiguous slides are made, the spread overlap. This creates an overall dose increase



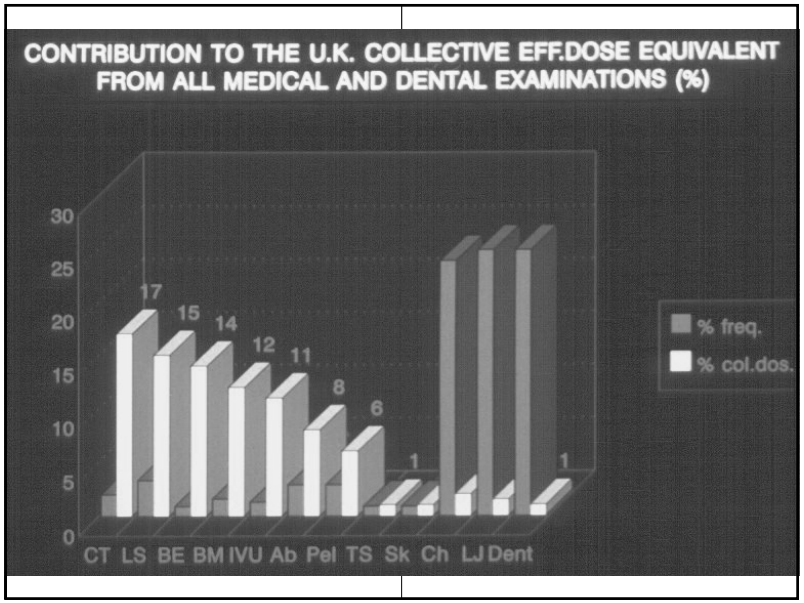
Computed Tomography

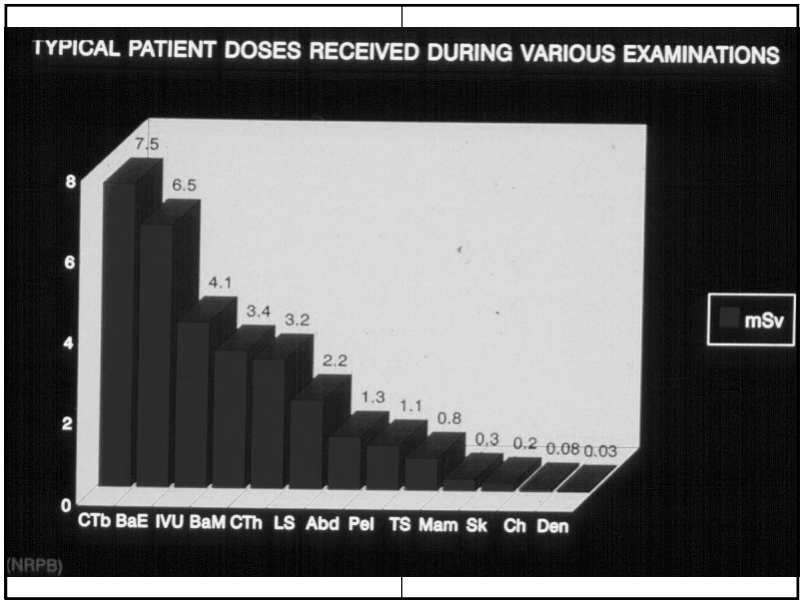
Dose Index - CTDI

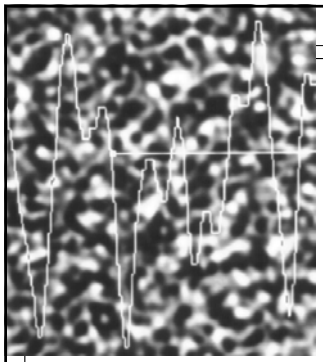
Originally the integration is over 14T

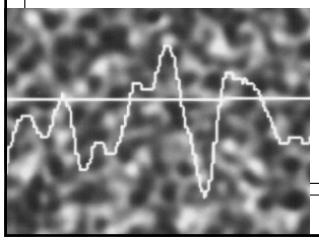
$$CTDI = \frac{1}{n.T}\int_{z1}^{z2} D(z)dz \quad \textbf{mGy}$$

z1,z2 = the limits of integration
D(z) = the single slice dose profile
T = the nominal slice thickness in cm
n = the number of slices irradiated simult





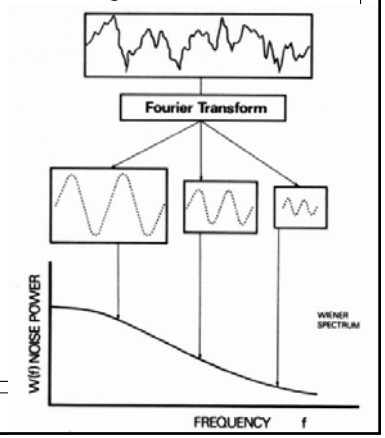


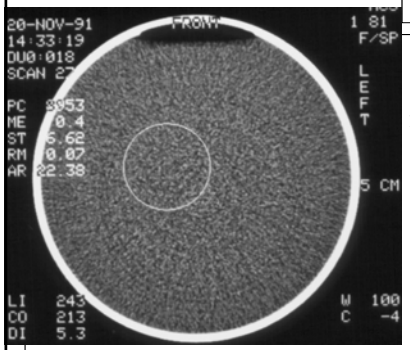


CT Image - Noise

-Enlarged image of water phantom (high and low noise level)

- Wiener noise spectrum calculation

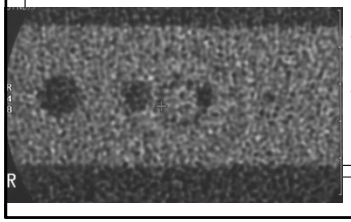




- CT noise as standard deviation of mean CT values (HU) varies most often between 1 and 10 HU

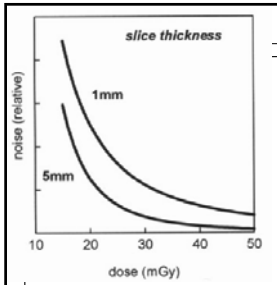
-Main limitation of contrast resol.

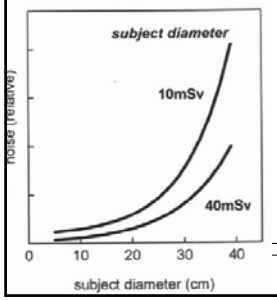
- Min. contrast > noise level (HU)



- Standard deviation (noise):

$$\sigma = [\Sigma(CT_i - CT_{mean})^2 / (n-1)]^{1/2}$$





CT Noise (σ) depends mainly on:

- Reconstruction algorithm and filter - C
- Pixel size (FOV / matrix size) - P
- Scan Dose - D
- Slice thickness - ST
- Object parameters - (μd)

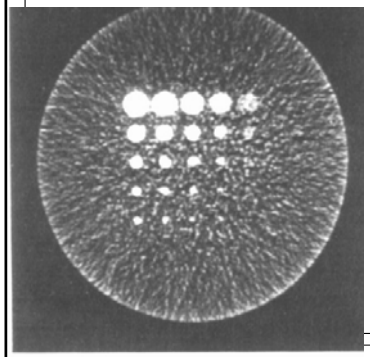
- Formula of Brooks & DiChiro for noise (σ) dependence:

$$\sigma \sim C\,[(e^{-\mu d}) / (D \times ST \times P^3)]^{1/2}$$

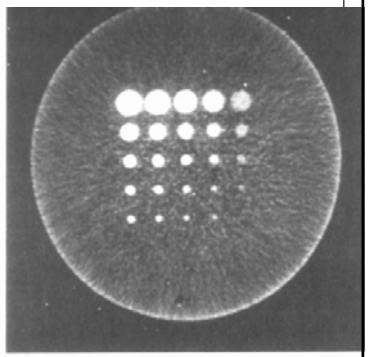
Basic Scanning parameters

- kV, mA (mAs) - the increase of these:

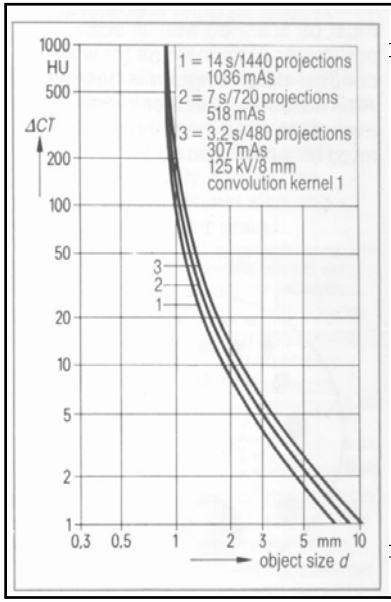
increases the contrast resolution (decreases the noise), hence the spatial resolution , but also increases the patient dose



10 mGy



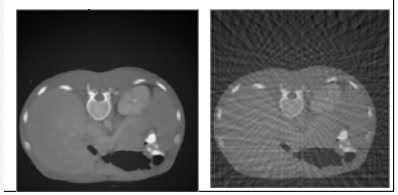
80 mGy



Basic Scanning parameters

- Scan time (number projections, sampling rate) - increase of these:

increases the resolution (both high contrast and low contrast), but also increases the patient dose



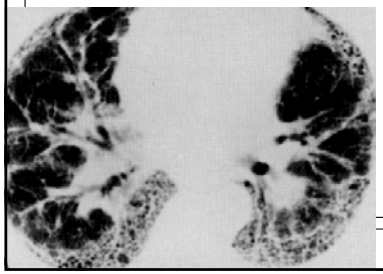
1024 projections

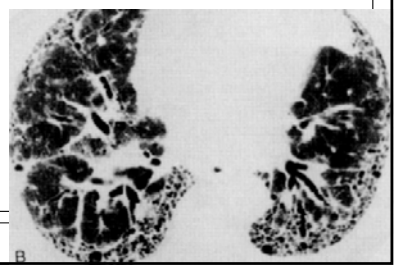
128 projections

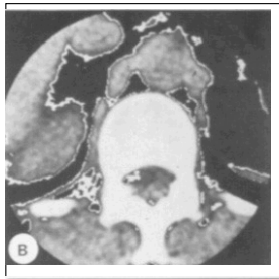
Basic Scanning parameters

- Slice thickens - its increase:

decreases the spatial resolution (both in X-Y and Z direction) and increases the patient dose (might increase slightly the contrast of large homogeneous objects)







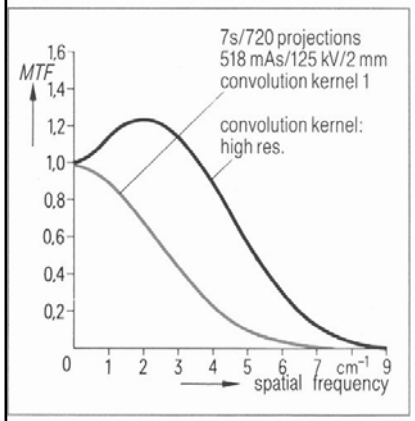
Basic Scanning parameters

Field of View (FOV)

Also Basic Imaging parameter.

FOV/matrix size = pixel size

Its increase leads to increase of pixel size, hence decrease of spatial resolution.

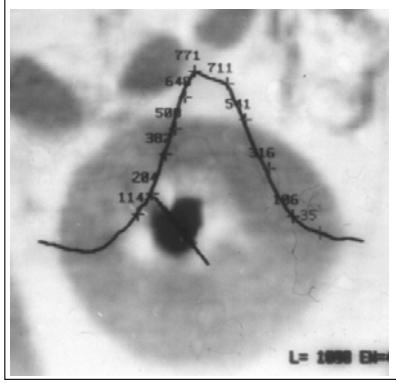


Basic Imaging parameters

Reconstruction algorithm

also Basic Scanning parameter

The HF algorithm increase high spatial resolution, but also increases the noise, hence decreases the contrast resolution (all algorithms are different for Head and Body)



All scanning and imaging parameters (but mainly the algorithm) influence the image contour spread.

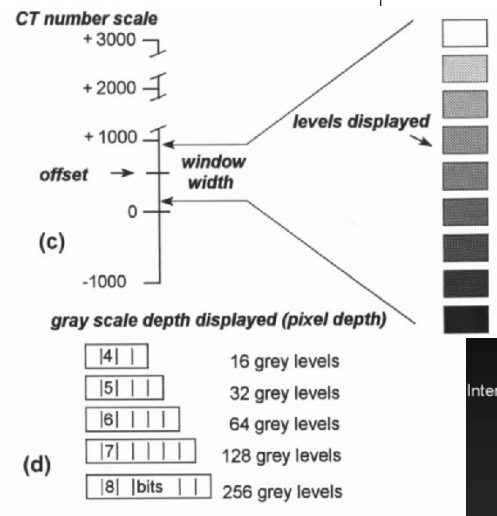
This spread is one of the main reasons for un-precise densitometry of small objects.

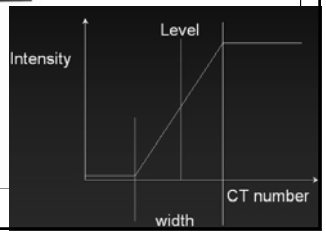
Basic Imaging parameters

Image filter

The hard filter (HF) filter increase high spatial resolution, but also increases the noise, hence decreases the contrast resolution. The filter is often mistaken with the algorithm, although they lead to similar effects, the filter is applied after the algorithm. The filter can be changed for any image. Soft filtration (LF) increases contrast resolution.

Basic Imaging parameters



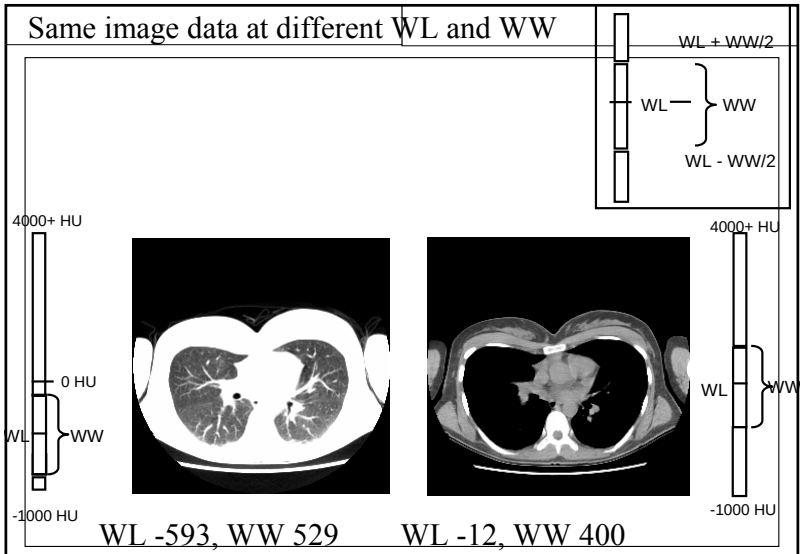


Windowing (WW,WC)

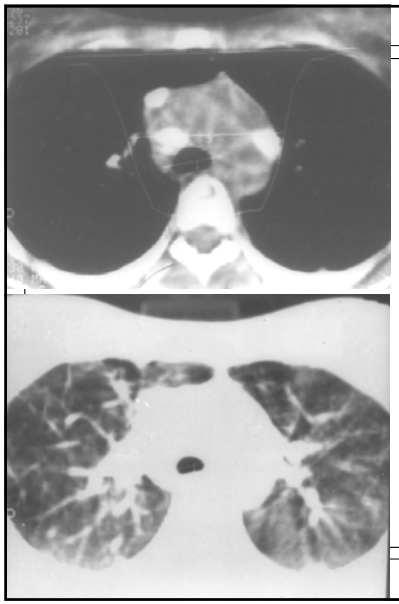
The only technique to allow visual use of the high CT contrast resolution.

Most important for image assessment.

Same image data at different WL and WW



WL -593, WW 529 WL -12, WW 400

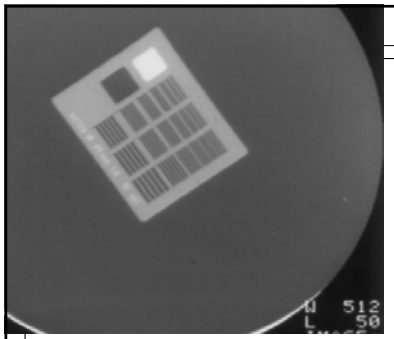


WC changes not only the visualised tissue range, but also the size of the displayed objects.

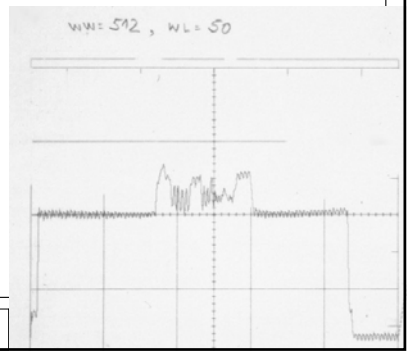
WW and WC should be adjusted for each object (tissue).

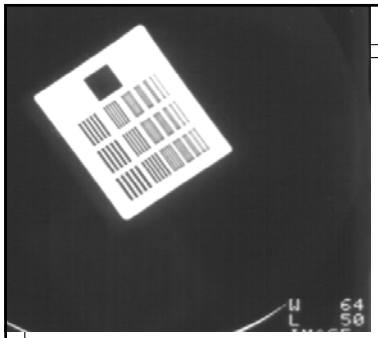
Setting WW, WC

1. Measure the object density (HU)
2. Set the WC at this HU value
3. Start with narrow WW, enlarging it until best display of the object and its surroundings.

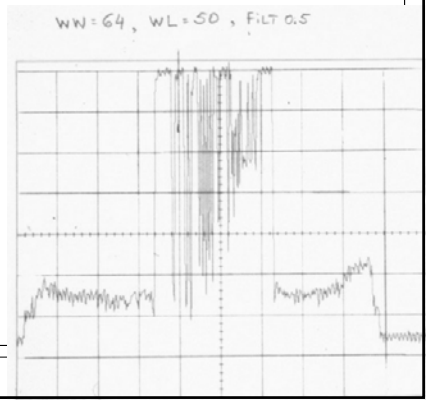


WW changes the image contrast - its amplitude is well seen from the video signal of the displayed image.





Narrow WW increases the contrast dynamic, what lead to visual increase of the spatial resolution.



WW, WC, and all image processing changes dramatically the visualisation.

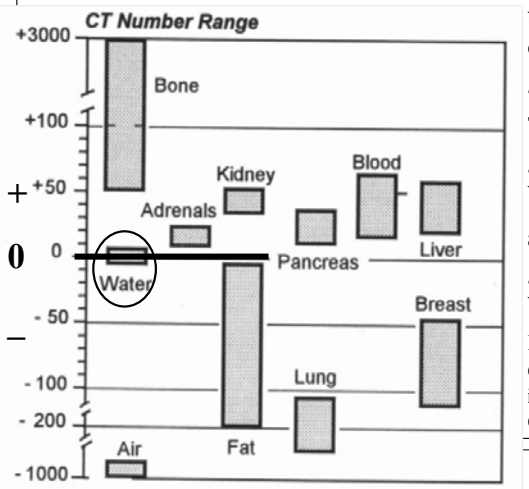
These should be recorded to assure repeatability of results

The calculated pixel values in the memory are not influenced by WW and WC. This is not true for some image processing

$CT_{number} = 1000 . (\mu_{tissue} - \mu_{water}) / \mu_{water}$

CT number (CT#, HU)

$CT_{number} = 1000 . (\mu_{tissue} - \mu_{water}) / \mu_{water}$

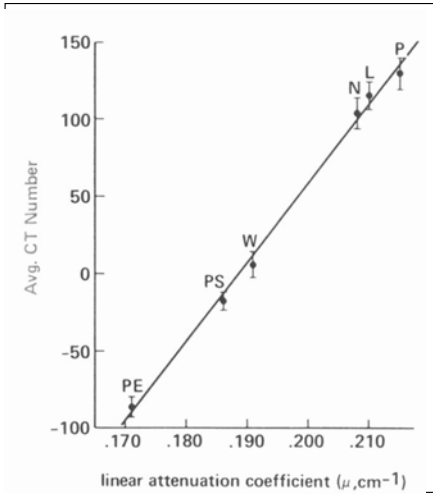


CT number represents the Lin. Attn. Coef. **μ** of an object, related to **μ water** (CT water=0)

This way objects with **μ object > μ water** will have **+CT number**

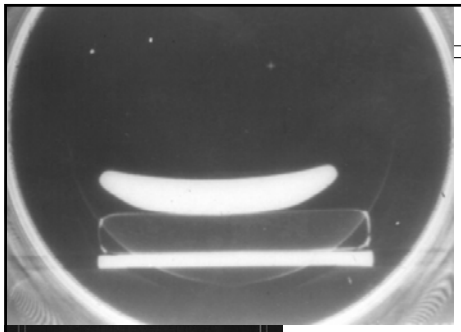
and objects with **μ object < μ water** will have **-CT number**

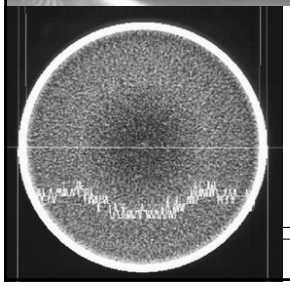
Body organs on CT image differ not only with their image, but also with their CT # (CT densitometry)



Precise adjustment of the CT Densitometry is performed during the CT calibration (usually made by the CT engineer).

Additional adjustments are made automatically every day (or before each scanning period)





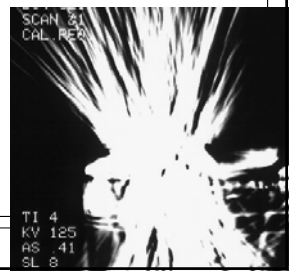
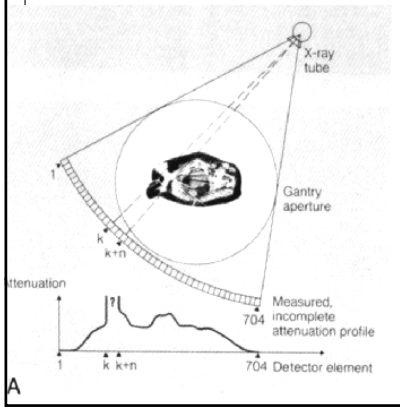
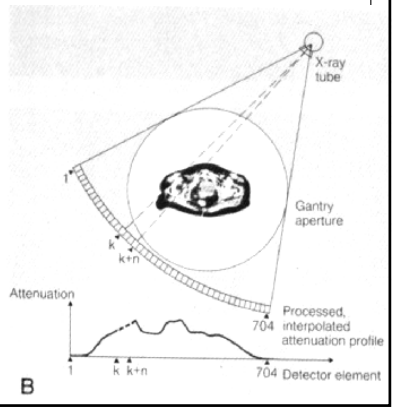


Image Artefacts

Beam hardening artefact exists in all CT scans (with different intensity)

Correction of beam hardening (metal artefact) with interpolation of the attenuation profile (from Felsenberg)





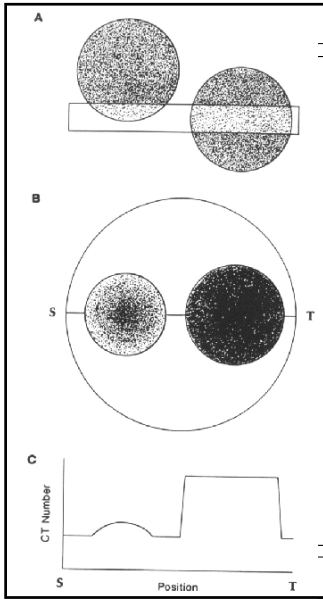


Image artefacts

Partial volume - present in all scans

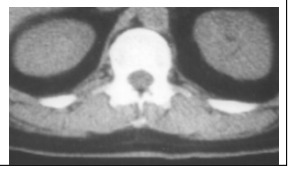


Image artefacts

Motion artefact