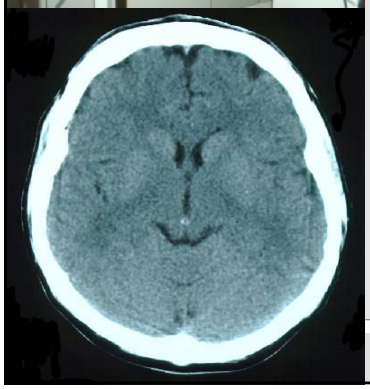
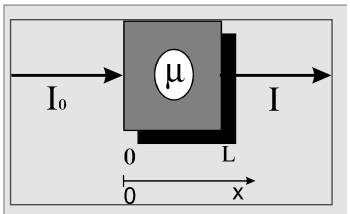


Basic principles of CT scanners and image reconstruction

CT scanning and imaging parameters

Dr Slavik Tabakov

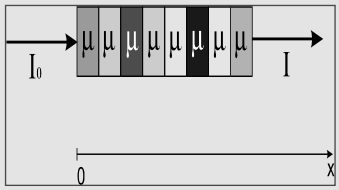
Dept. Medical Eng. and Physics
King's College London

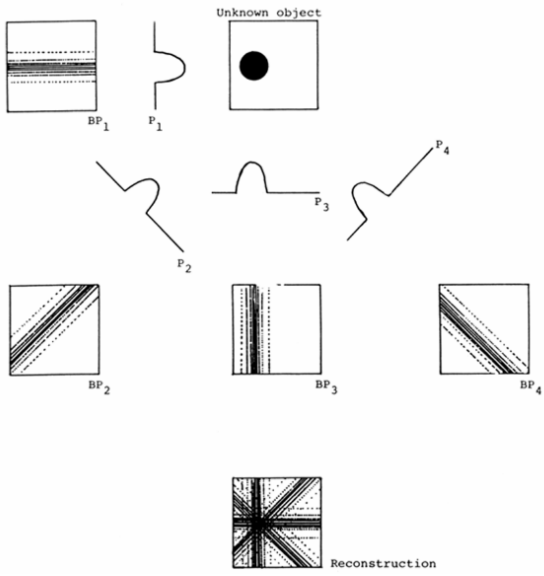


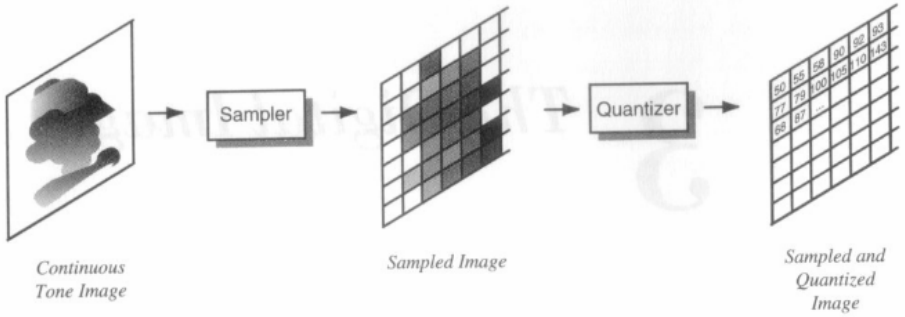
? How to visualise all parts of a "closed" 3D object ?

-voxel: elementary volume

-pixel: element of the image

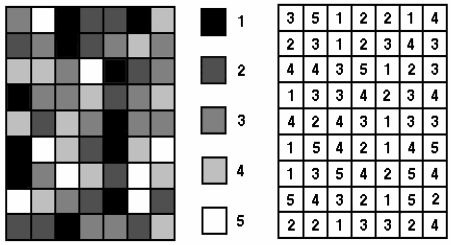


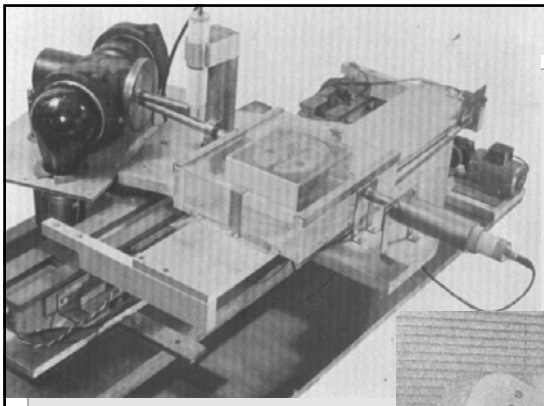




2 types of images :

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



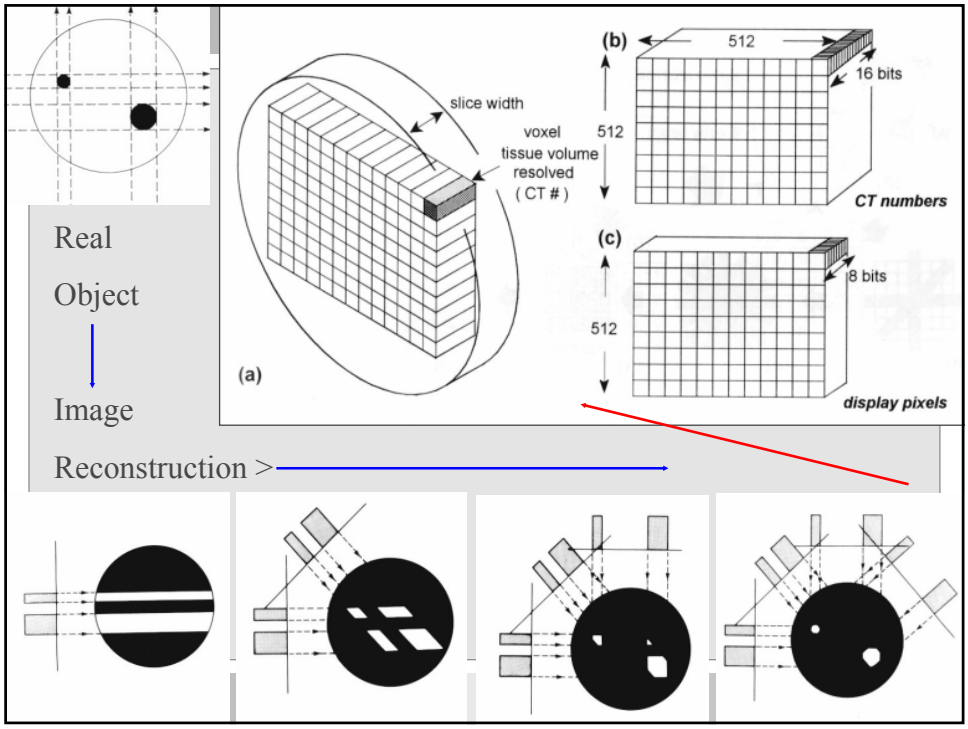


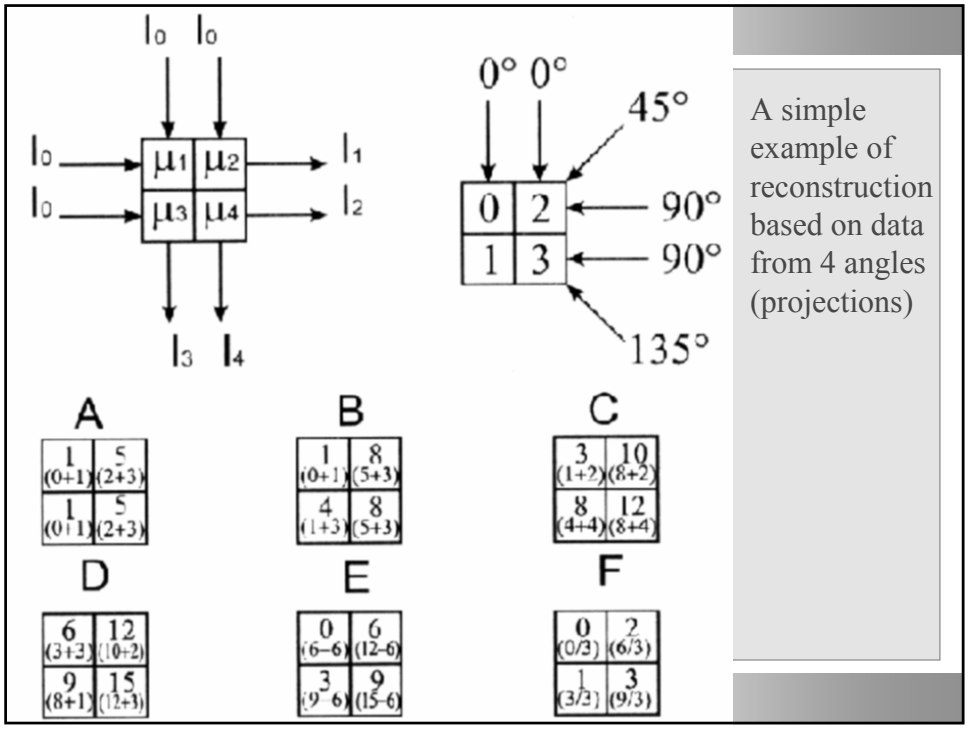
Mathematical theory (Radon,1917)

Applications in astronomy (1950)

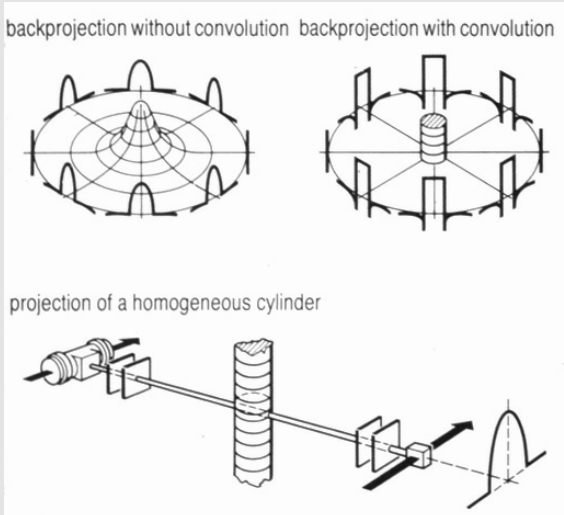
First CT experimental set (1960)

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



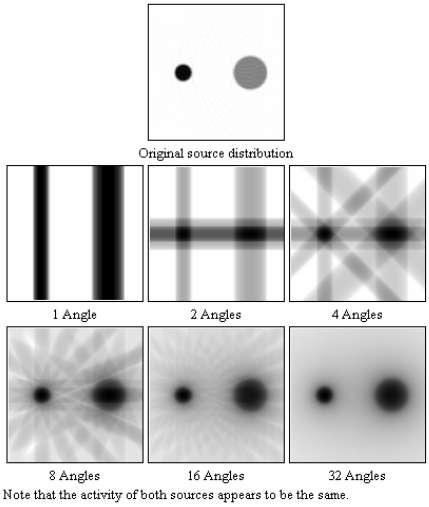


A simple example of reconstruction based on data from 4 angles (projections)



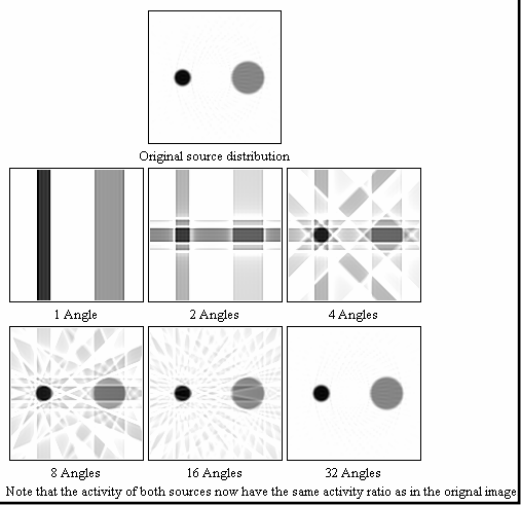
CT image is a product of complex calculation based on backprojection reconstruction algorithm.

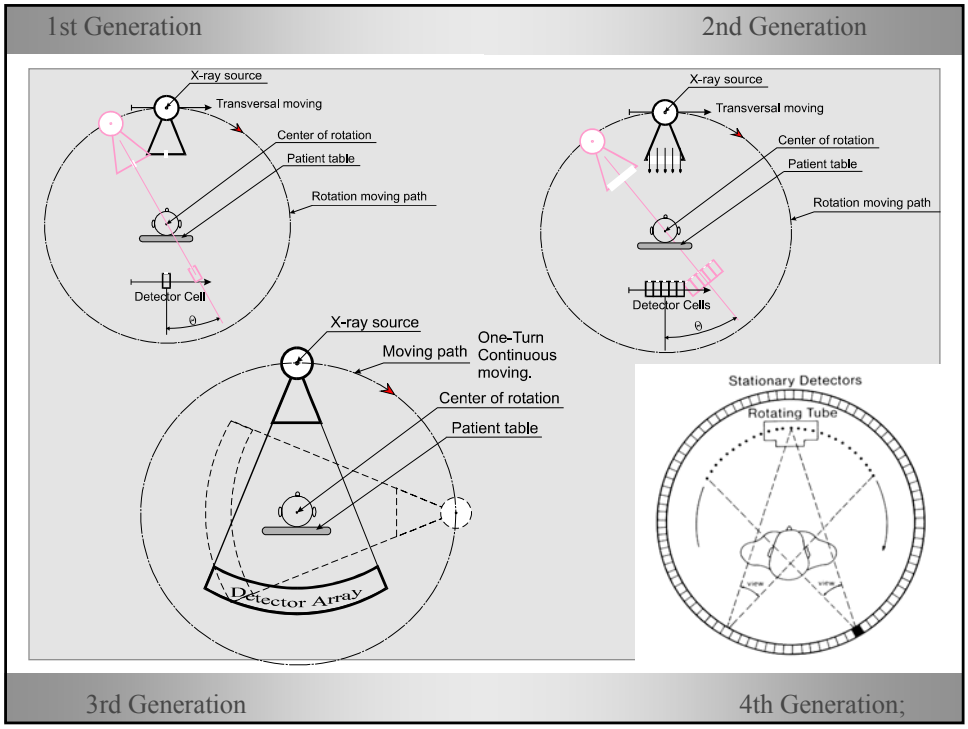
The CT image is not a real "shade" like in classical Radiography, but a picture which represents with some probability (>95%) the similarity between the real object and its calculated CT image.

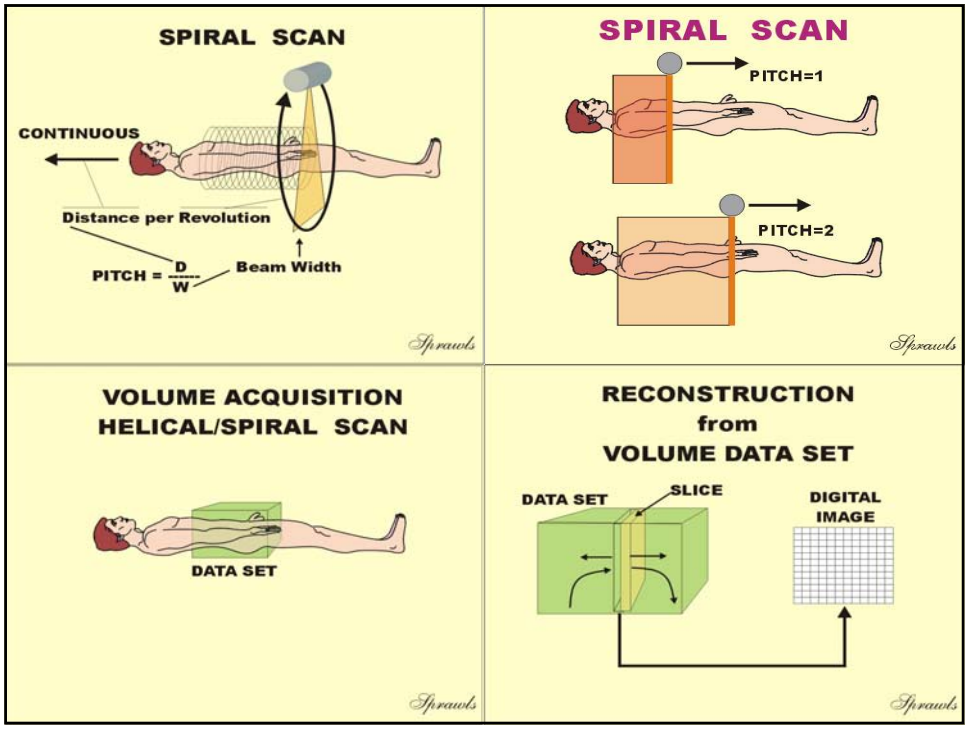


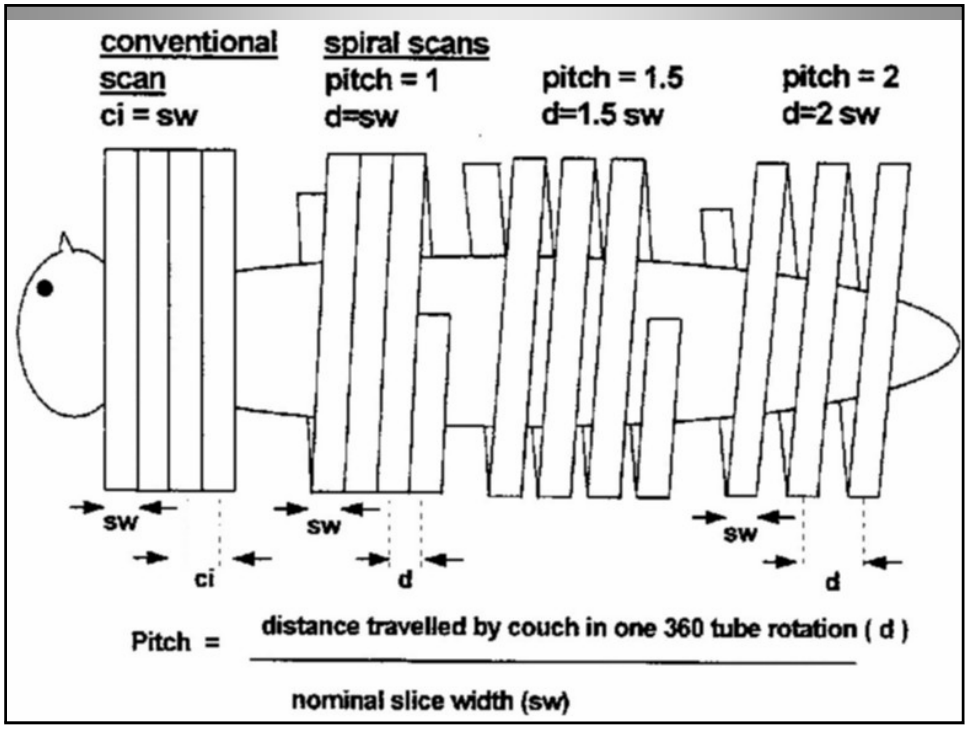
2. Filtered backprojection >>>

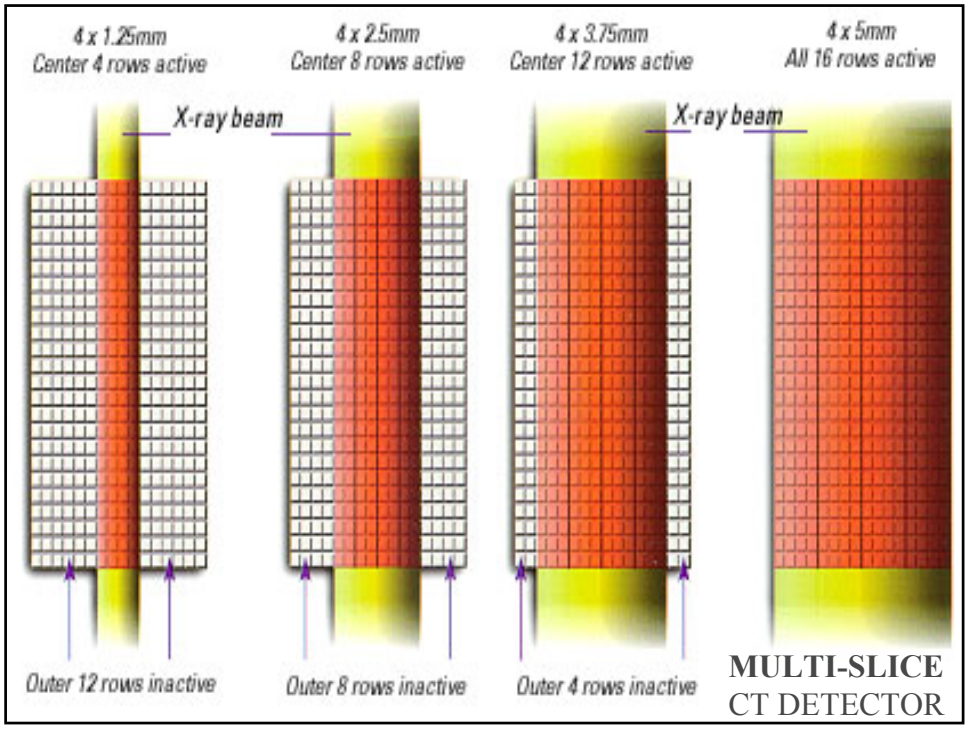
<<< 1. Simple (unfiltered) backprojection



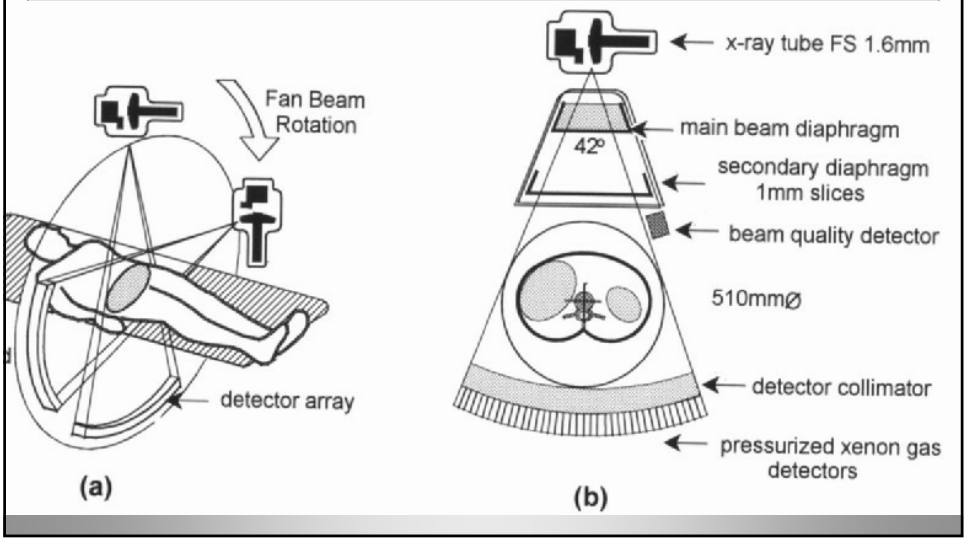

1st Generation
2nd Generation
X-ray source
Transversal moving
Center of rotation
Patient table
Rotation moving path
Detector Cell
θ
X-ray source
Transversal moving
Center of rotation
Patient table
Rotation moving path
Detector Cells
θ
X-ray source
Moving path
One-Turn
Continuous
moving.
Center of rotation
Patient table
Detector Array
Stationary Detectors
Rotating Tube
3rd Generation
4th Generation;

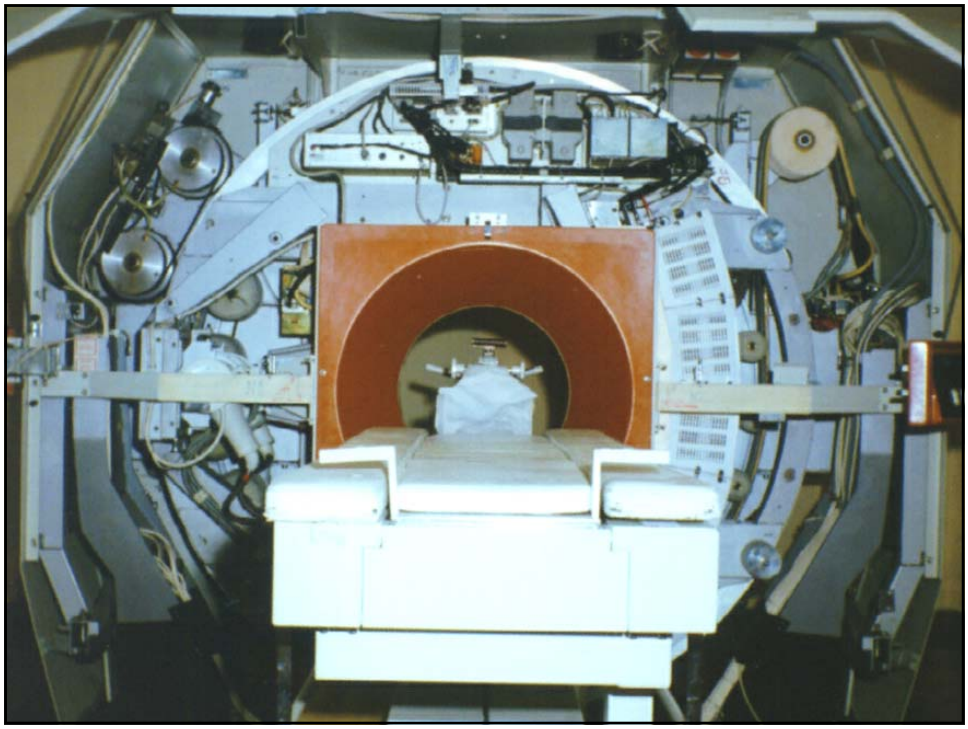

SPIRAL SCAN
CONTINUOUS
Distance per Revolution
PITCH = D / W
Beam Width
Sprawls
SPIRAL SCAN
PITCH=1
PITCH=2
Sprawls
VOLUME ACQUISITION
HELICAL/SPIRAL SCAN
DATA SET
Sprawls
RECONSTRUCTION
from
VOLUME DATA SET
DATA SET
SLICE
DIGITAL
IMAGE
Sprawls


conventional scan
ci = sw
spiral scans
pitch = 1
d=sw
pitch = 1.5
d=1.5 sw
pitch = 2
d=2 sw
sw
sw
sw
ci
d
d
Pitch = distance travelled by couch in one 360 tube rotation (d)
nominal slice width (sw)

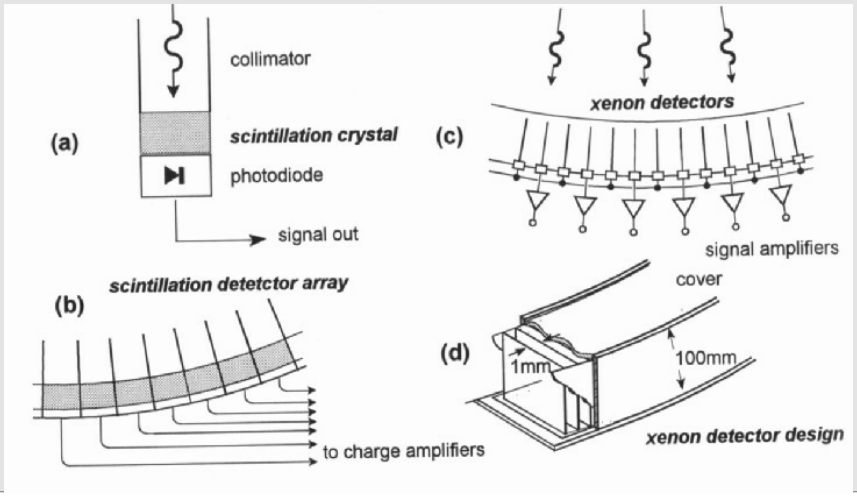

4 x 1.25mm
Center 4 rows active
4 x 2.5mm
Center 8 rows active
4 x 3.75mm
Center 12 rows active
4 x 5mm
All 16 rows active
X-ray beam
X-ray beam
Outer 12 rows inactive
Outer 8 rows inactive
Outer 4 rows inactive
MULTI-SLICE
CT DETECTOR

Typical III generation CT scanner

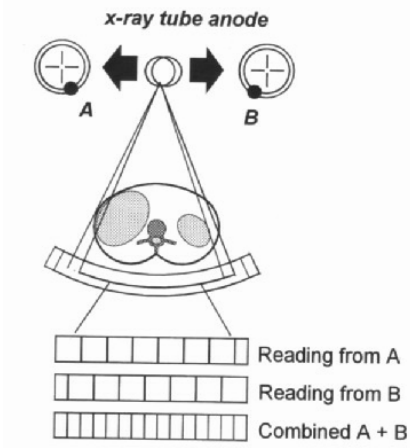


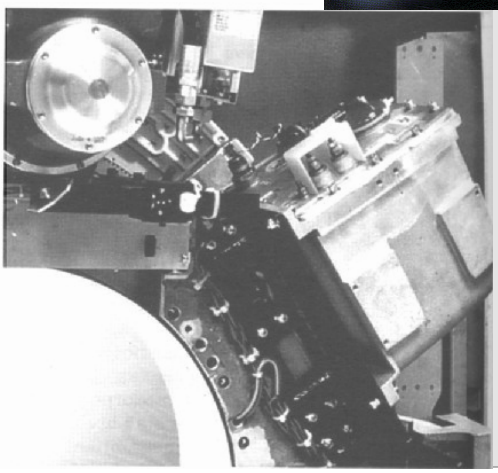


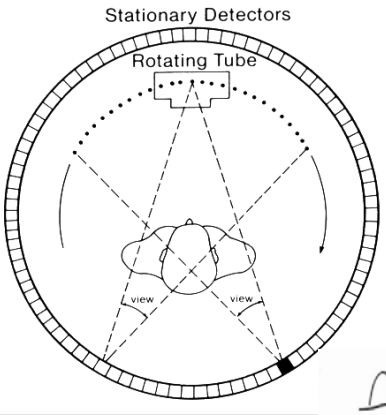
CT detectors development - solid state and gas (Xe) filled
Increased detector efficiency and new (III) generation scanners



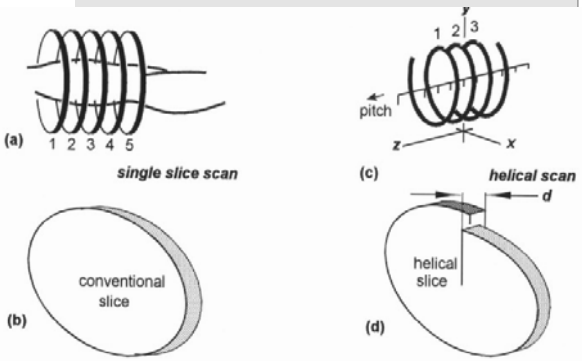
Increase spatial resolution using powerful new
X-ray tubes with flying focal spot technology

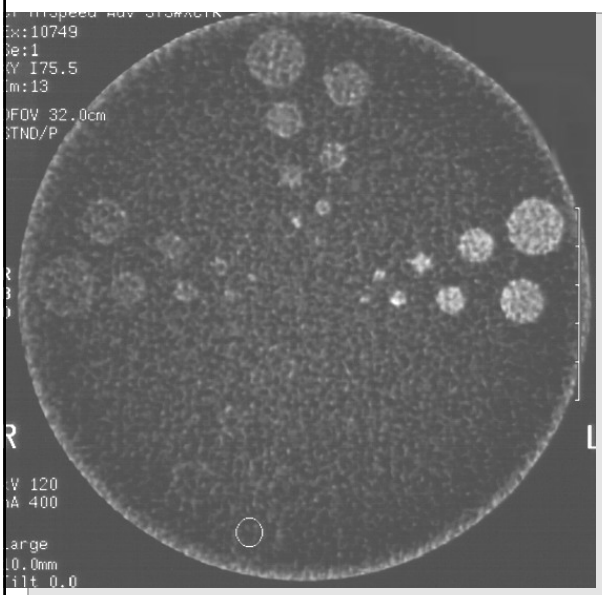






Typical IV generation CT scanner and Spiral (Helical) CT scanning



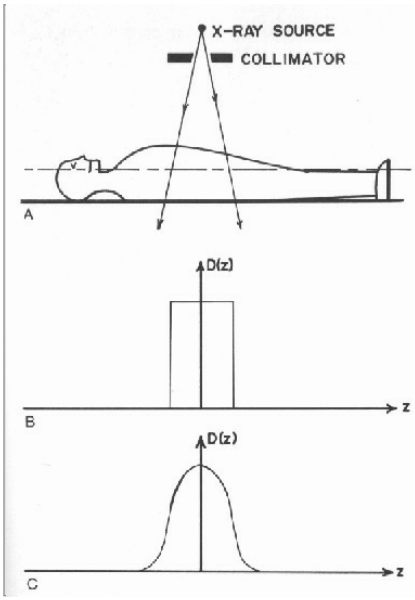


CT contrast resolution is one of the dramatic advantages of the method, compared with classical radiography.

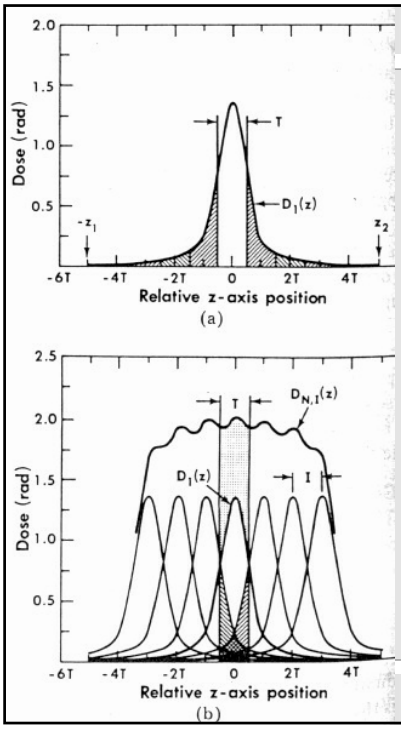
CT discriminates density difference ~ 0,25-0.5 % while normal Radiography discriminates ~ 10%

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

9



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

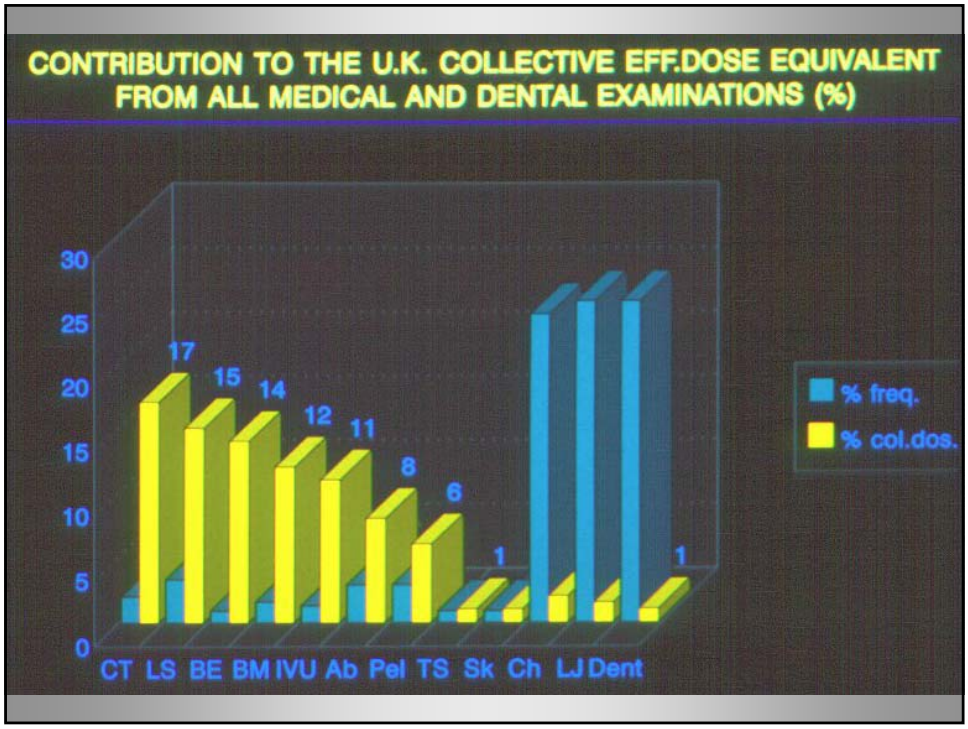
$$\mathrm{CTDI} = \frac{1}{n.T}\int_{z1}^{z2} D(z)dz \quad \mathbf{mGy}$$

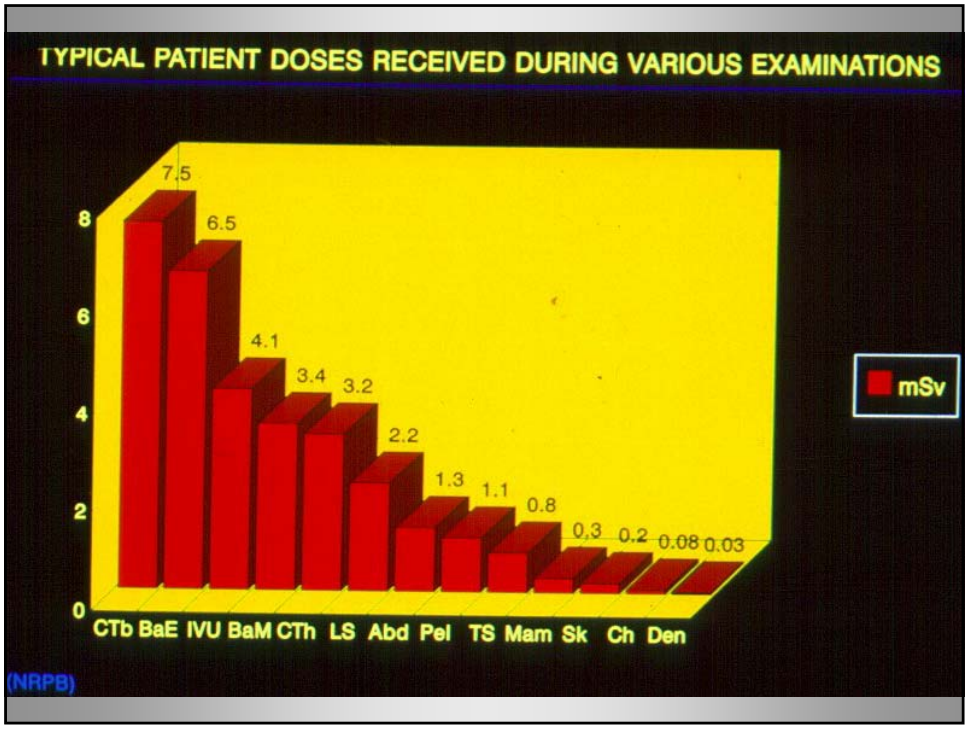
n = the number of slices irradiated simultaneously

T = the nominal slice thickness in cm

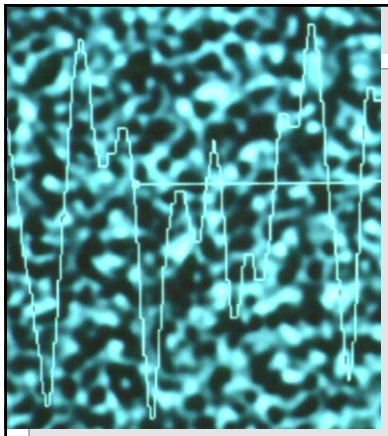
D(z) = the single slice dose profile

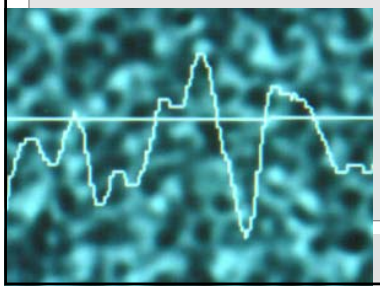
z1,z2 = the limits of integration


CONTRIBUTION TO THE U.K. COLLECTIVE EFF.DOSE EQUIVALENT FROM ALL MEDICAL AND DENTAL EXAMINATIONS (%)
30
25
20
15
10
5
0
17
15
14
12
11
8
6
1
1
% freq.
% col.dos.
CT LS BE BM IVU Ab Pel TS Sk Ch LJ Dent


TYPICAL PATIENT DOSES RECEIVED DURING VARIOUS EXAMINATIONS
8
6
4
2
0
7.5
6.5
4.1
3.4
3.2
2.2
1.3
1.1
0.8
0.3
0.2
0.08
0.03
mSv
CTb BaE IVU BaM CTh LS Abd Pel TS Mam Sk Ch Den
(NRPB)

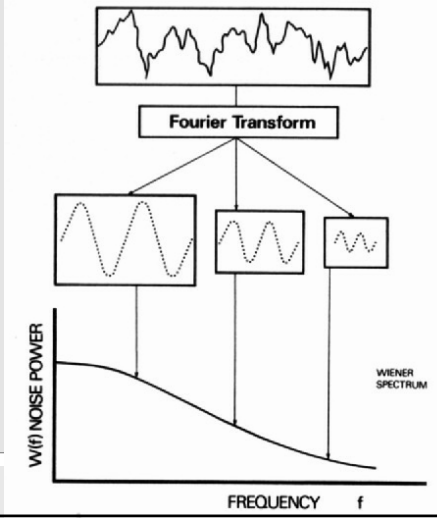
CT Image - Noise

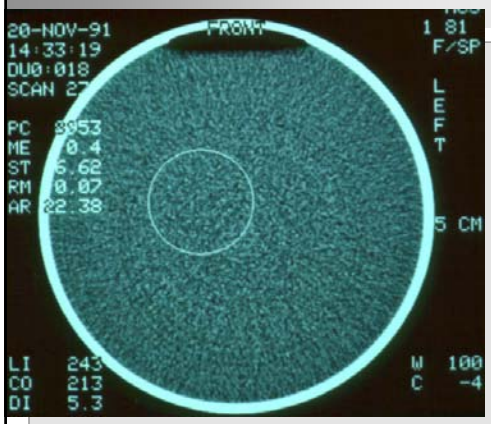


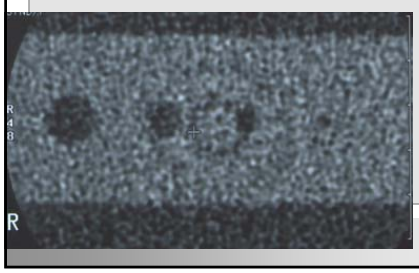


-Enlarged image of water phantom (high and low noise level)

- Wiener noise spectrum calculation







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

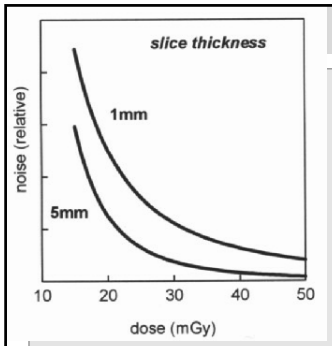
-Main limitation of contrast resol.

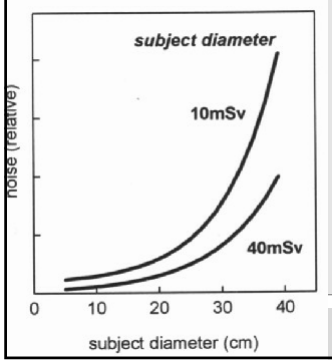
- Min. contrast > noise level (HU)

- Standard deviation (noise):

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

12





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)

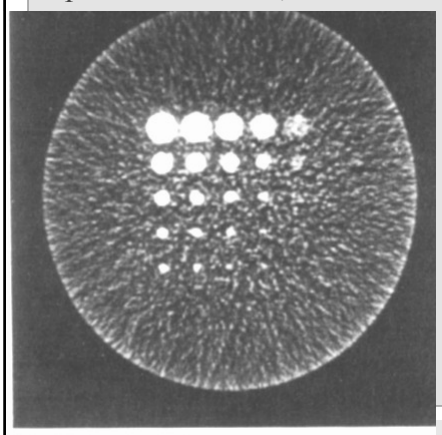
- Formula of Brooks & DiChiro for noise (σ) dependence:

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

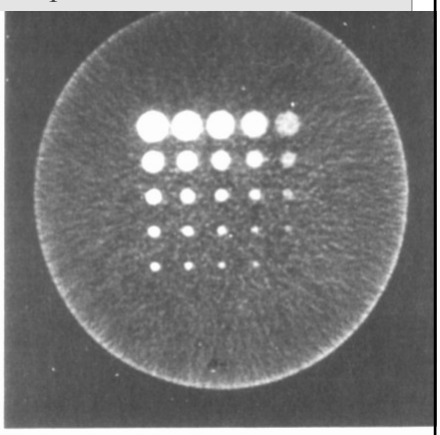
Basic Scanning parameters

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

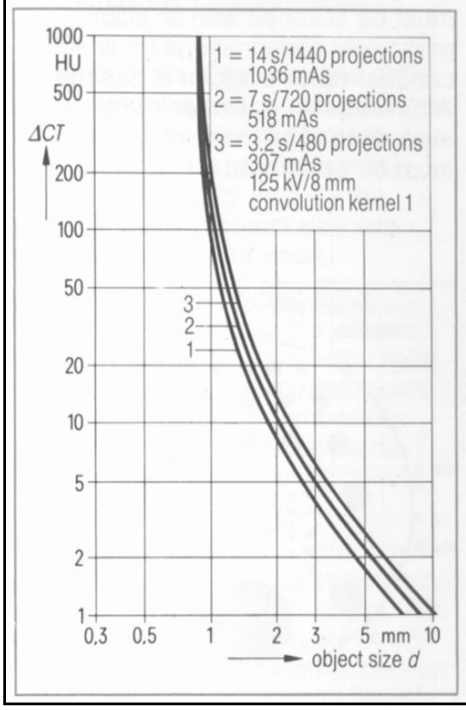
increases the contrast resolution (decreases the noise), hence the spatial resolution , but also increases the patient dose



10 mGy



80 mGy



Basic Scanning parameters

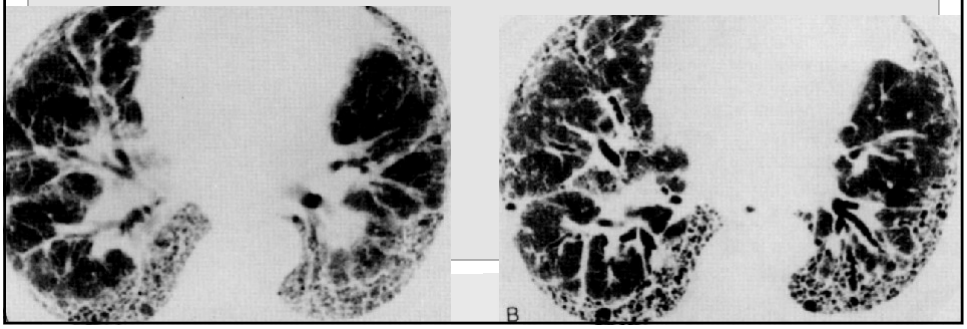
- Scan time (number projections) - the increase of these:

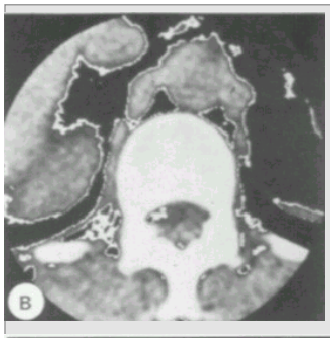
increases the resolution (both high contrast and low contrast), but also increases the patient dose

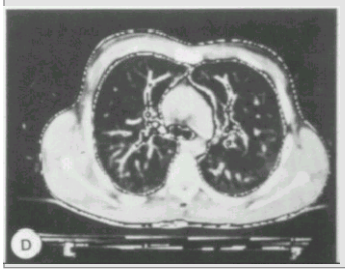
Basic Scanning parameters

- 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)







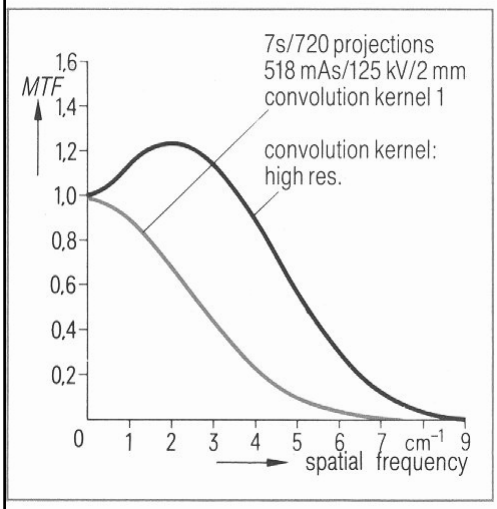
Basic Scanning parameters

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.

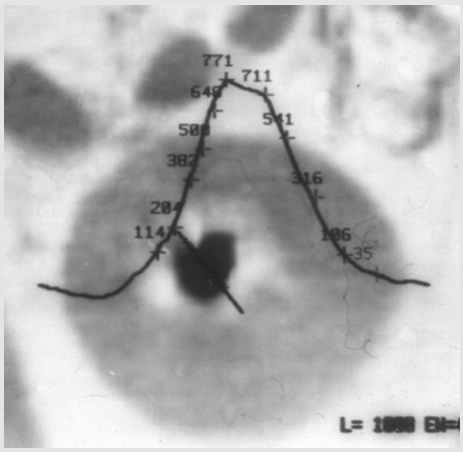


Basic Imaging parameters

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)



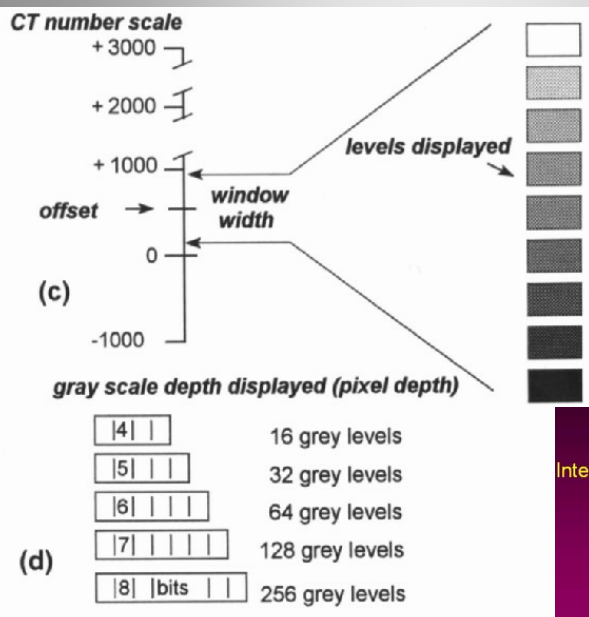
All scanning and imaging parameters (but mainly the algorithm) influence the image contour spread.

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

Basic Imaging parameters

Image filter

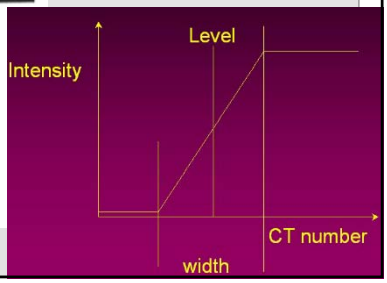
The hard filter (HF) filter increase high spatial resolution, but also increases the noise, hence decreases the contrast resolution. The filter is often mistaken with the algorithm, although they lead to similar effects, the filter is applied after the algorithm. The filter can be changed for any image.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

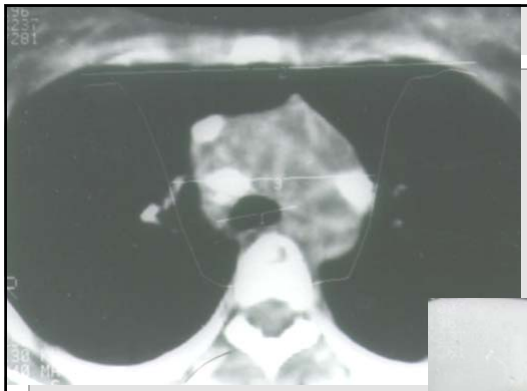


Windowing (WW,WC)

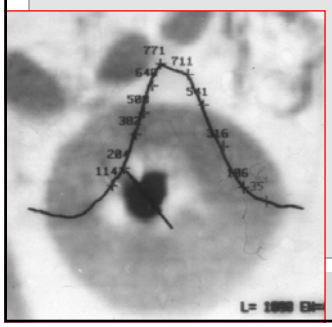
The only technique to allow visual use of the high CT contrast resolution.

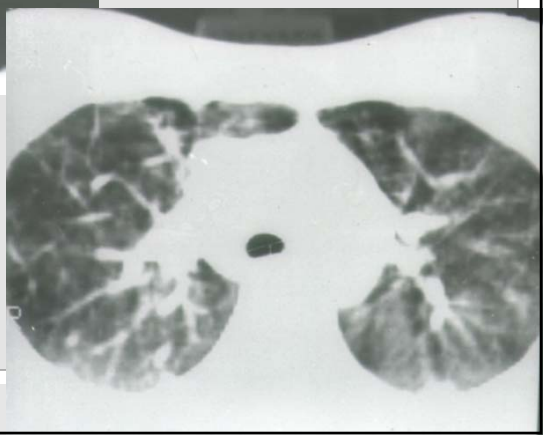
Most important for image assessment.

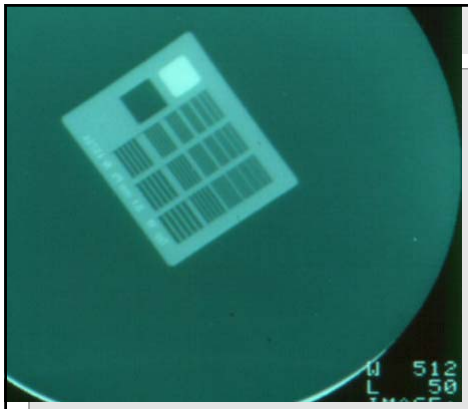




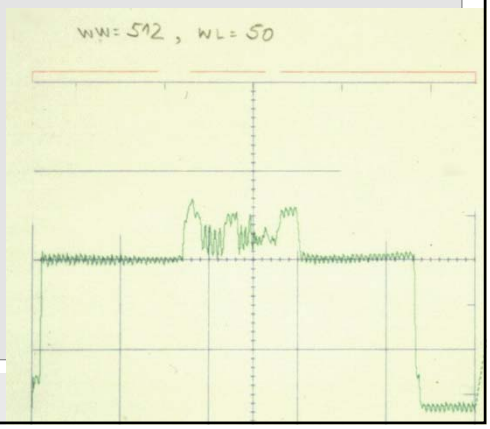
WC changes not only the visualised tissue range, but also the size of the displayed objects (structures)

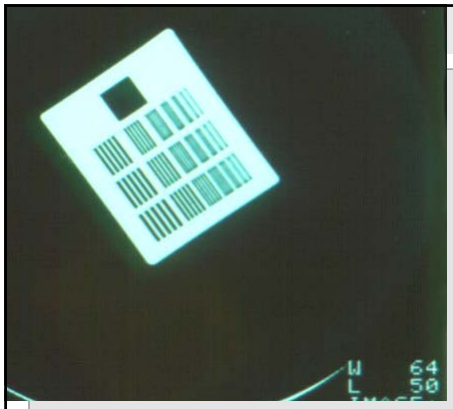




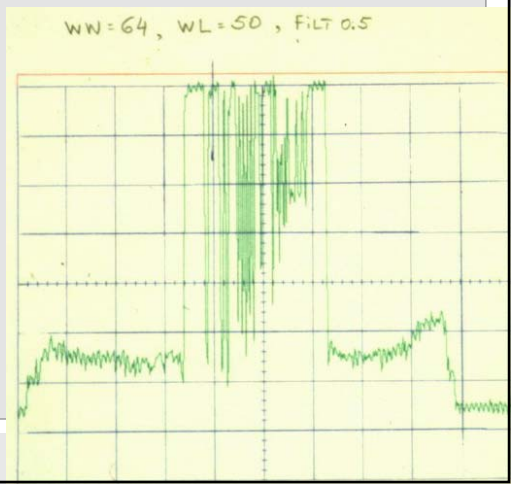


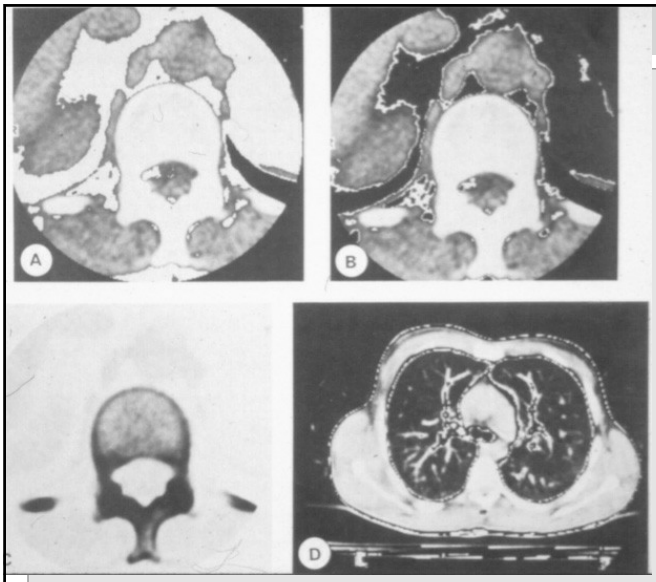
WW changes the image contrast - its amplitude is well seen from the video signal of the displayed image.





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



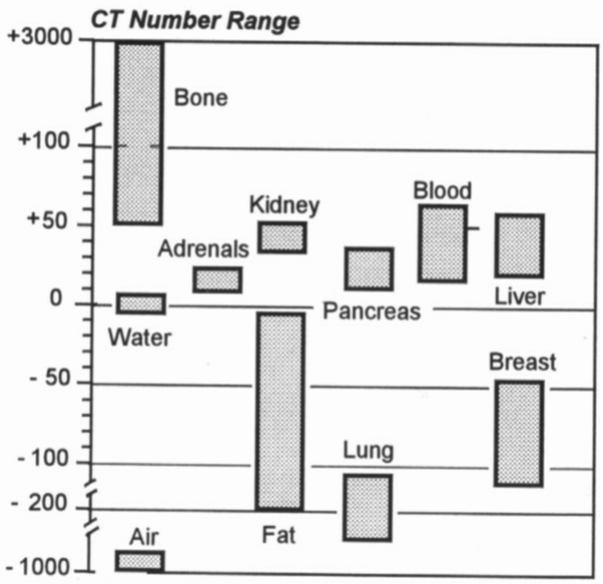


WW, WC, and all image processing changes dramatically the visualisation.

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

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



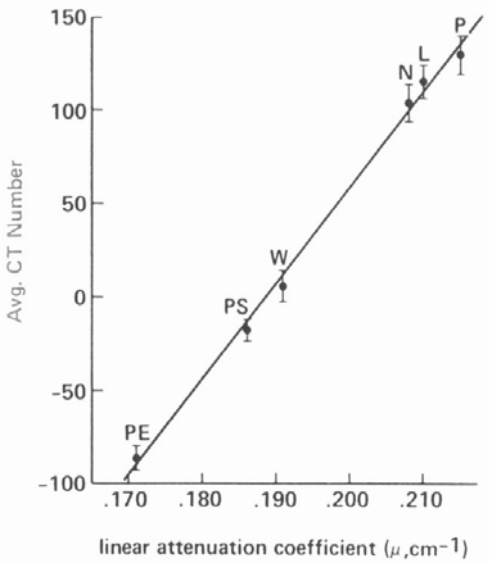
WW and WC should be adjusted for each object (tissue).

Setting WW, WC

1. Measure the object density (HU).

2. Set the WC at this value

3. Start with narrow WW, enlarging it until best display of the object and its surroundings.



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)

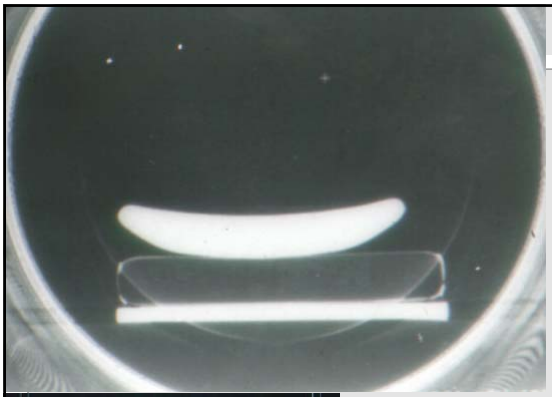
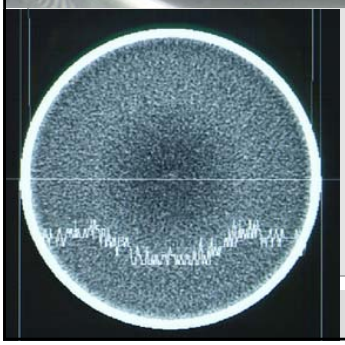
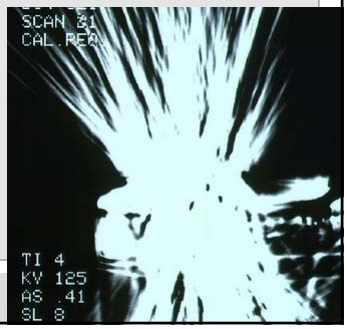


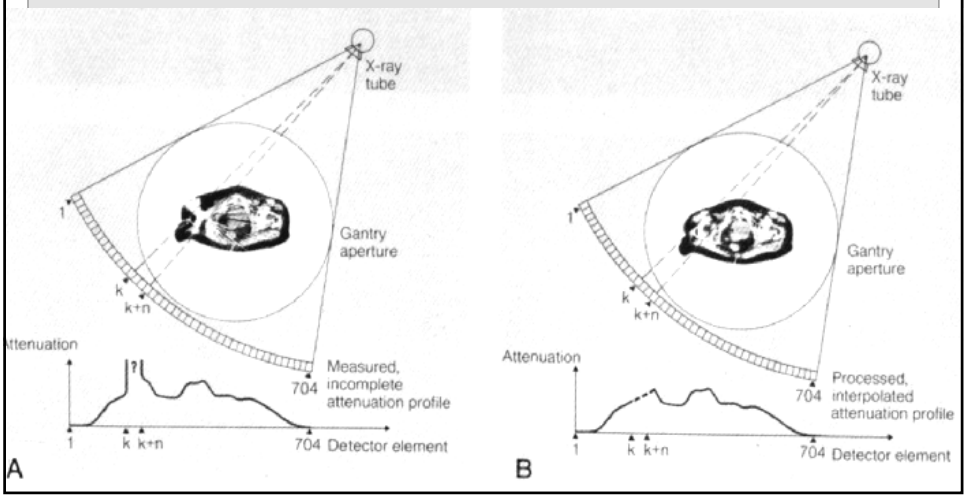
Image Artefacts

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)



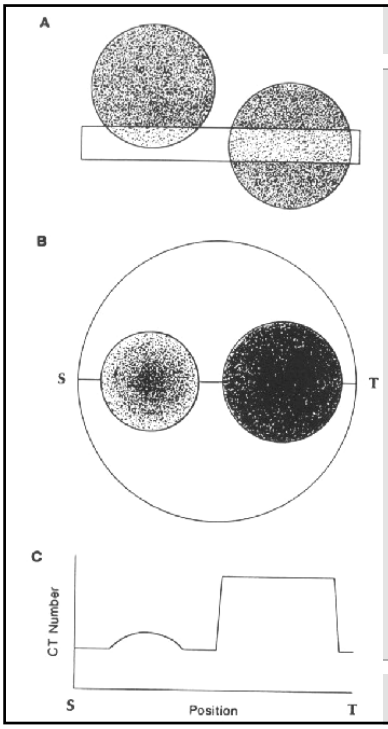


Image artefacts

Partial volume - present in all scans

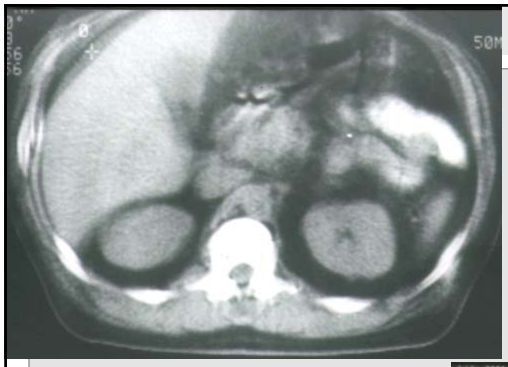
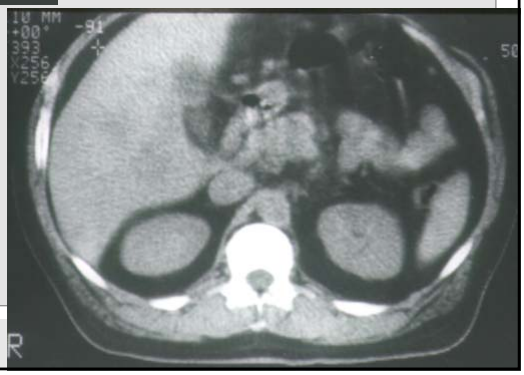


Image artefacts

Motion artefact



Contrast and Resolution of various X-ray detectors and methods

Contrast:

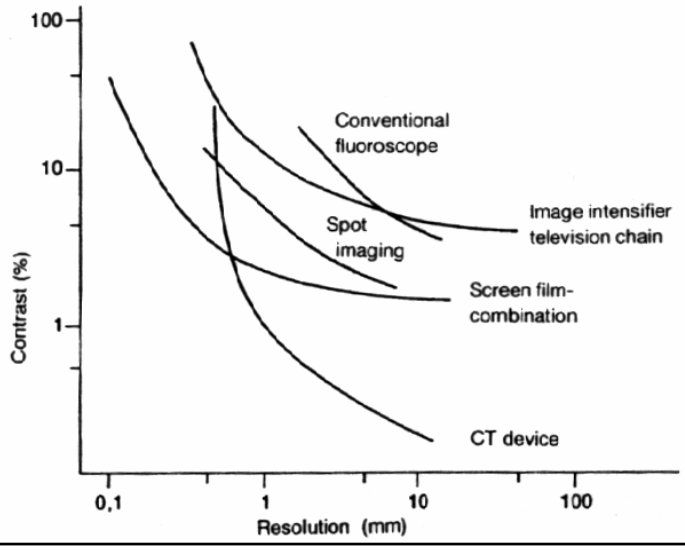
1. CT
2. Film
3. Fluo

Resolution:

1.Film

2.Fluo

3.CT



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