*«All bodies are transparent to this agent... For brevity's sake I shall use the expression ´rays´ and to distinguish them from others of this name I shall call them ´X-rays´»*

Wilhelm Röntgen

# FYS 4250
## Lecture 6



# Electromagnetic Spectrum

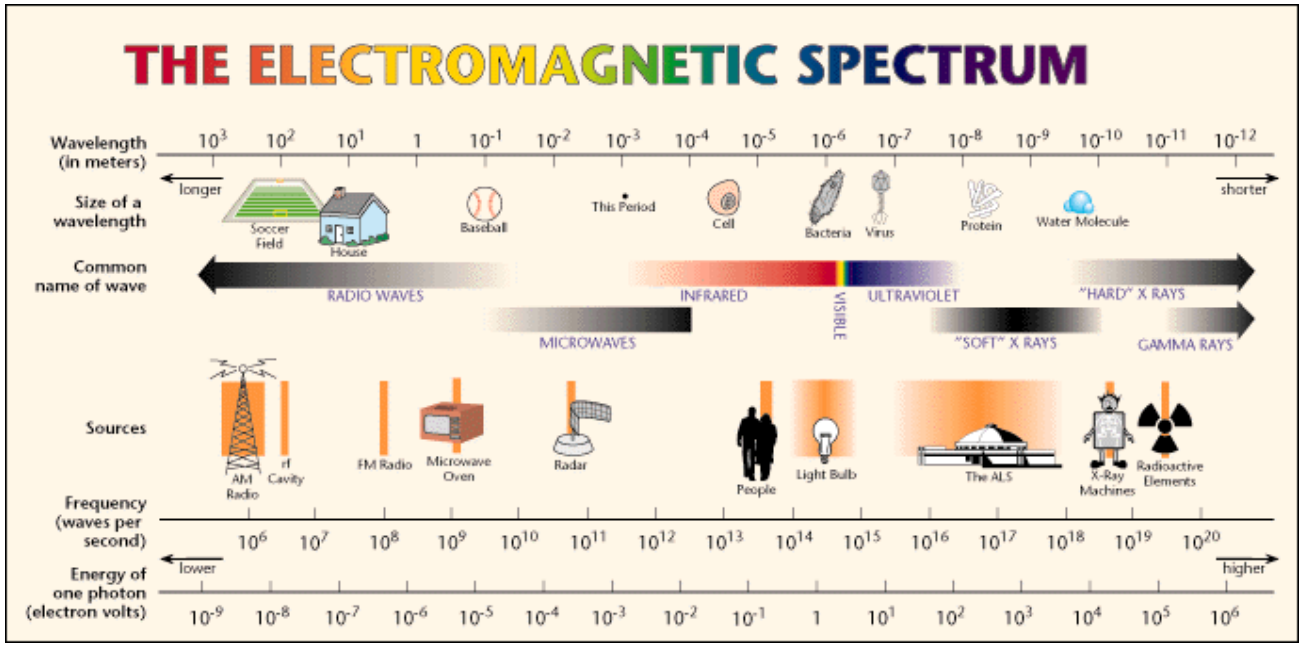



Image: www.lbl.gov

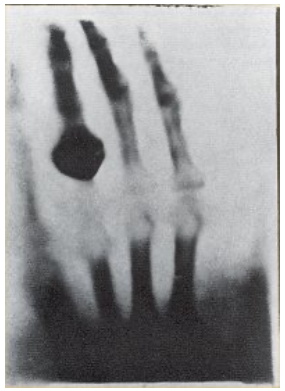
# Spatial resolution of imaging modalities

**TABLE 1-1 THE LIMITING SPATIAL RESOLUTIONS OF VARIOUS MEDICAL IMAGING MODALITIES. THE RESOLUTION LEVELS ACHIEVED IN *TYPICAL* CLINICAL USAGE OF THE MODALITY ARE LISTED**

| MODALITY | SPATIAL RESOLUTION (mm) | COMMENTS |
|---|---|---|
| Screen film radiography | 0.08 | Limited by focal spot size and detector resolution |
| Digital radiography | 0.17 | Limited by size of detector elements and focal spot size |
| Fluoroscopy | 0.125 | Limited by detector resolution and focal spot size |
| Screen film mammography | 0.03 | Highest resolution modality in radiology, limited by same factors as in screen film radiography |
| Digital mammography | 0.05–0.10 | Limited by same factors as digital radiography |
| Computed tomography | 0.3 | About ½ mm pixels |
| Nuclear medicine planar imaging | 2.5 (detector face), 5 (10 cm from detector) | Spatial resolution degrades substantially with distance from detector |
| Single photon emission computed tomography | 7 | Spatial resolution worst towards the center of cross-sectional image slice |
| Positron emission tomography | 5 | Better spatial resolution than the other nuclear imaging modalities |
| Magnetic resonance imaging | 1.0 | Resolution can improve at higher magnetic fields |
| Ultrasound imaging (5 MHz) | 0.3 | Limited by wavelength of sound |

Bushberg et al, The essential physics of medical imaging

# X-rays





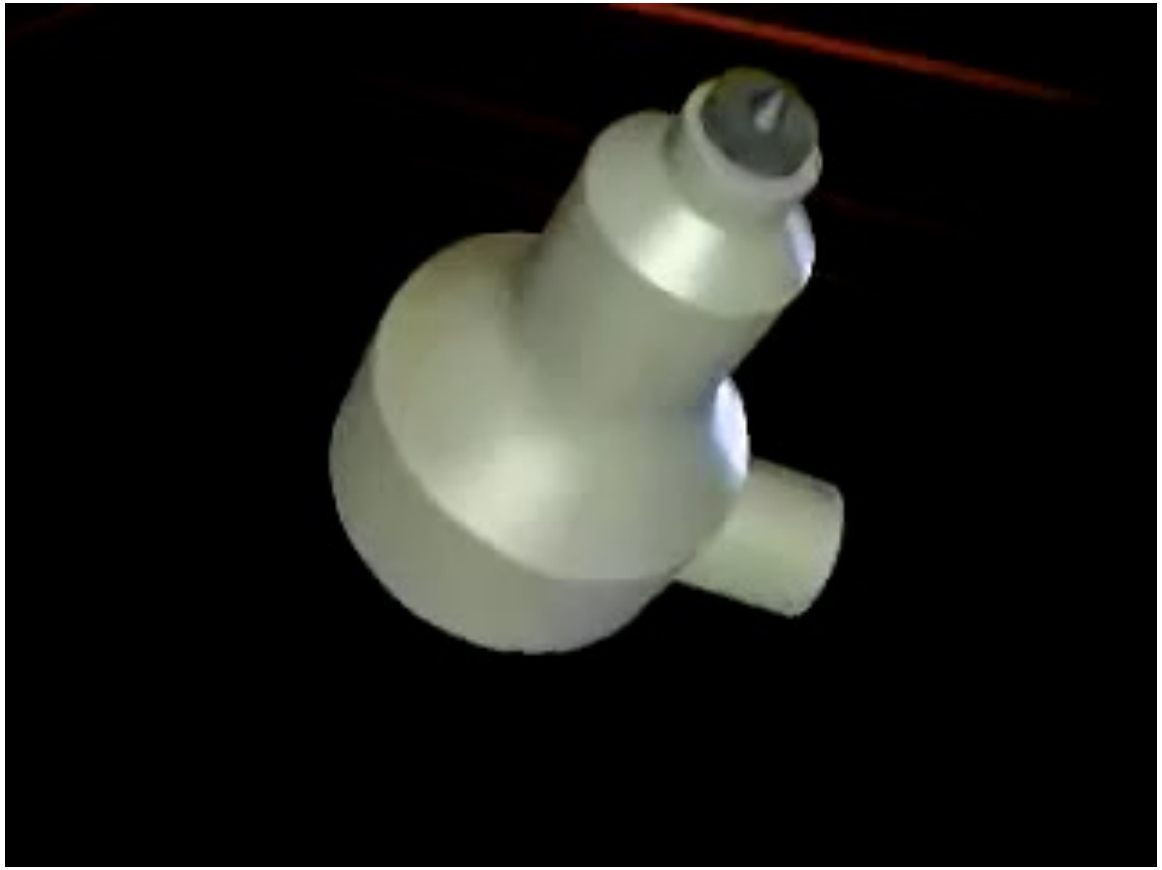
Discovered by Röntgen in 1895

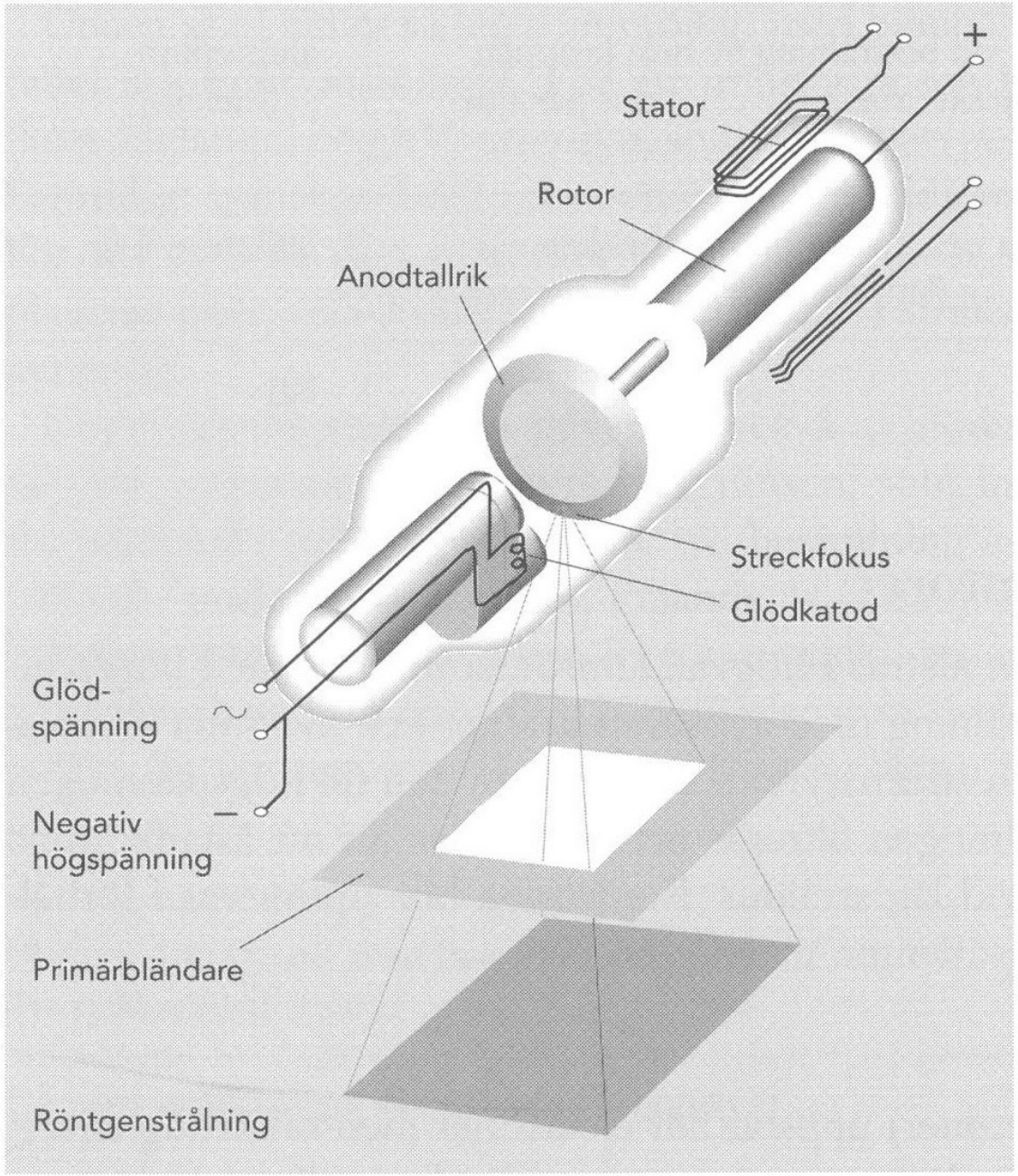
Incredible fast spread, first x-ray tube in Norway in 1897

X-rays are generated by bombarding electrons on a metallic anode

Wavelength is depending on the metal, X-rays from a Cu-anode is 12.398 keV

# X-ray system



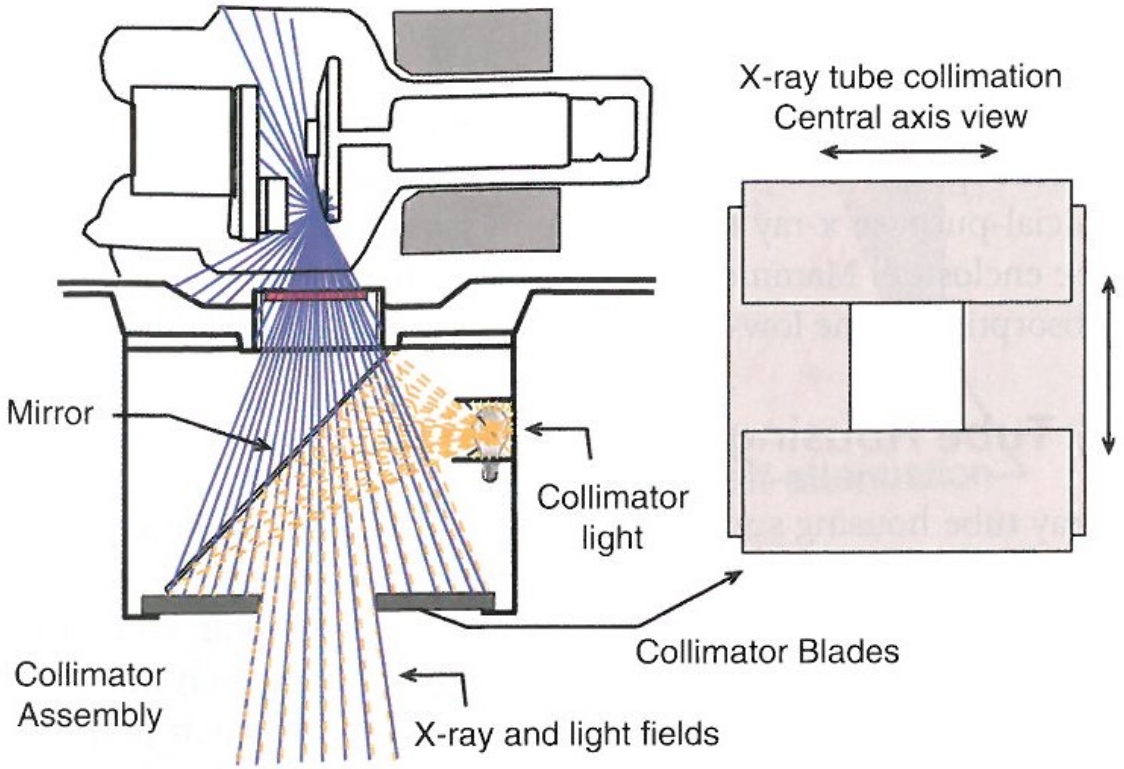



Figur 8:15. Röntgenrör med roterande anod, som drivs med ett roterande magnetfält.

# X-ray tube

What is the minimum wavelength?
Electron hits target -> all kinetic energy converted into a photon
Work done on each electron when accelerated to the anode is eV
hf = eV
Fmax=eV/h
Thus Landa (min) = hc/eV

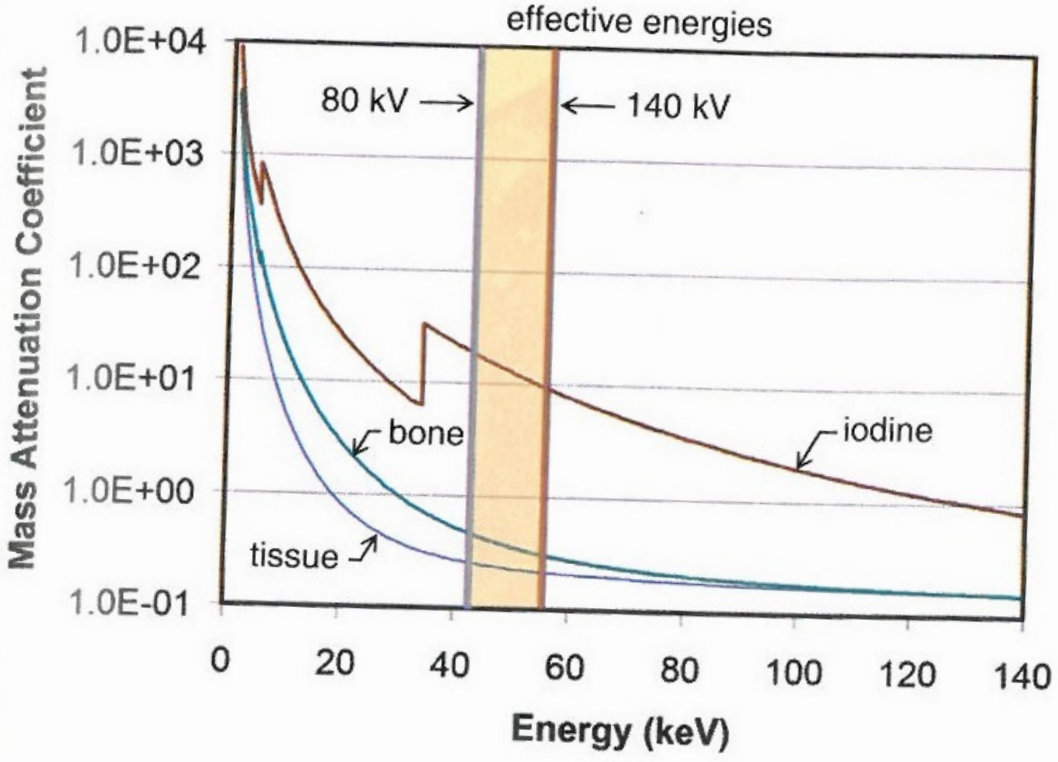
# X-ray tube




■ **FIGURE 6-19** The x-ray tube collimator assembly is attached to the housing at the tube port, typically on a collar that allows it to be rotated. A light source, positioned at a virtual focal spot location, illuminates the field from a 45-degree angle mirror. Lead collimator blades define both the x-ray and light fields.

Bushberg et al, The essential physics of medical imaging

# Attenuation as a function of x-ray energy




■ **FIGURE 10-20** This figure shows the mass attenuation coefficient for bone, tissue, and iodine as a function of x-ray energy. The attenuation coefficient is much greater for bone and greater still for 100% iodine. Typically, tissues have much lower concentrations of iodine. This figure illustrates the basis for contrast in CT between bone-, tissue-, and contrast-enhanced CT.

Bushberg et al, The essential physics of medical imaging

# Fotonspectrum X-ray tubes

Two mechanisms => Bremsstrahlung (deceleration of arriving electron) and characteristic radiation (electrons fill holes)
Characteristic radiation
58 keV = l-shell electron filling up hole in k-shell
68 = distant electron filling up hole in k-shell
Less than 1% of the energy supplied is converted into x-radiation, rest is heating of the object.

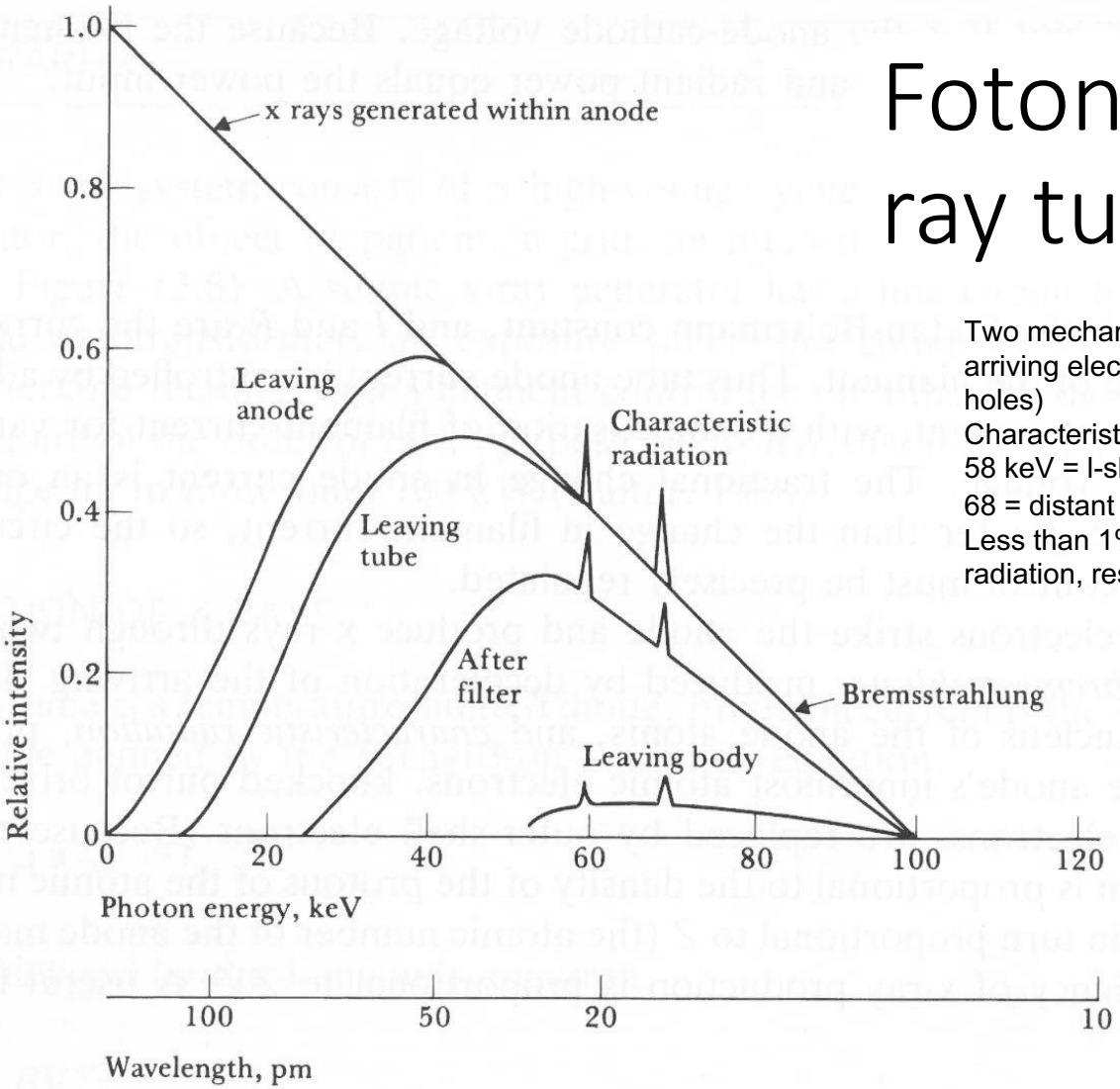



**Figure 12.9** The lowest-energy x rays are absorbed in the anode metal and the tube glass envelope. An Al filter further reduces the low-energy x rays that do not pass through the body and would just increase the patient dose. Only the higher-energy x rays are capable of penetrating the body and contributing to the film darkening required for a picture. Note that the average energy increases with the amount of filtration.

# Compton-spread: photon vs free electron

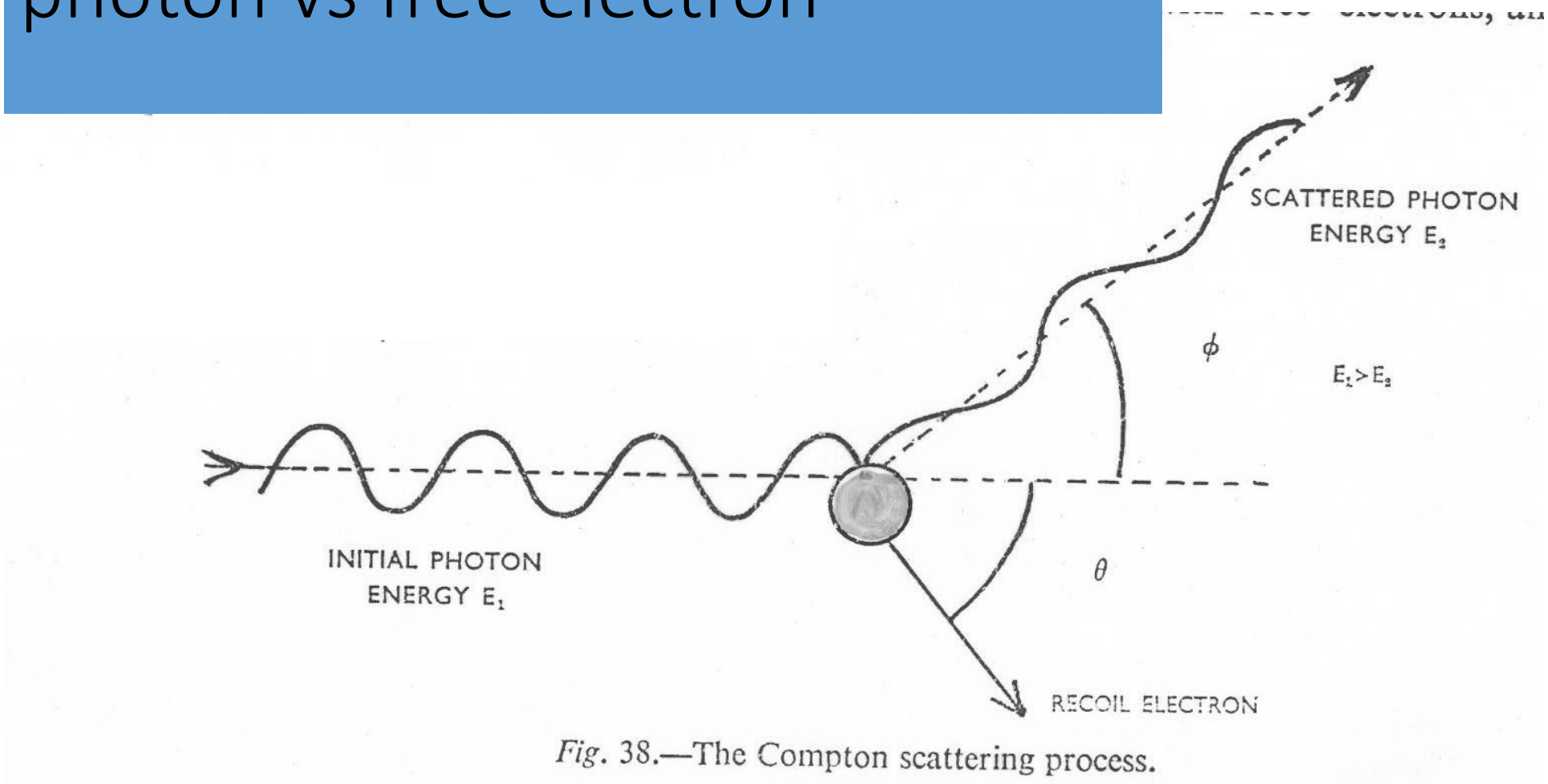



*Fig.* 38.—The Compton scattering process.

Secondary radiation,two mechanisms:
1. Photoelectric effect, characteristic radiation
2. Compton spread

Secondary radiation has always lower energy than the primary

# X-ray system

Electrons are colliding with metal, releasing energy. (Knocking electrons out of shell, outer electrons drop down to fill the vacancy -> emitting x-ray photons)
Electron beam is controlled by adjusting the filament current

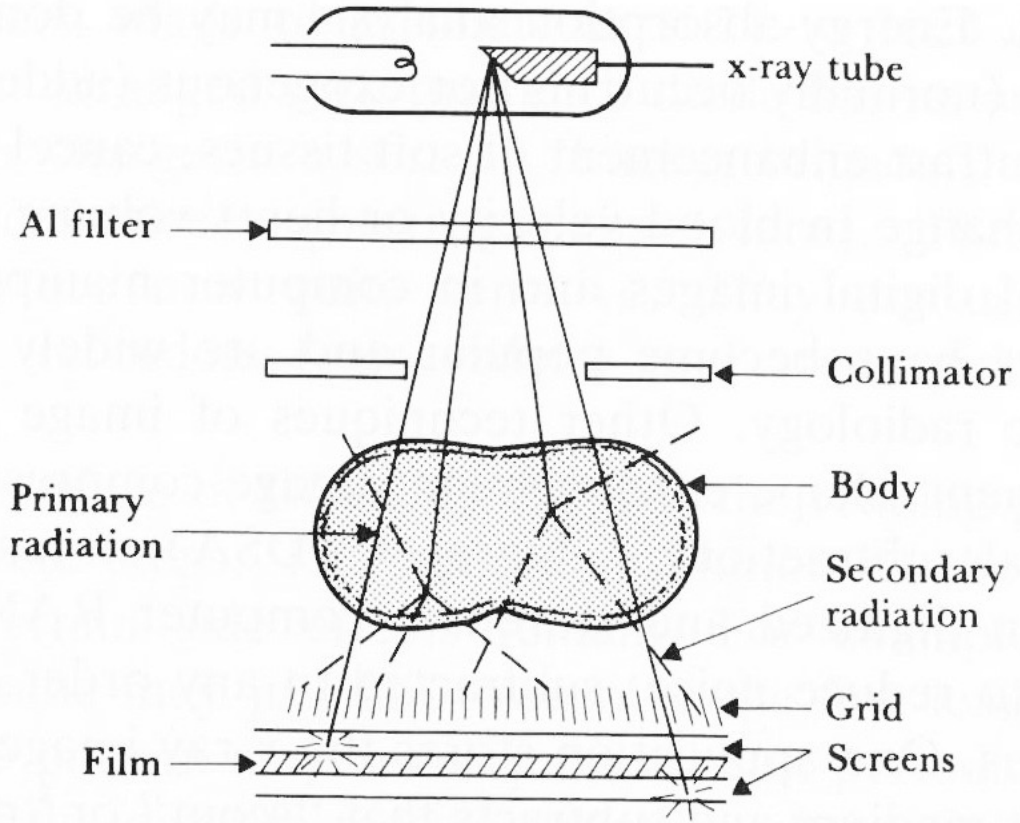
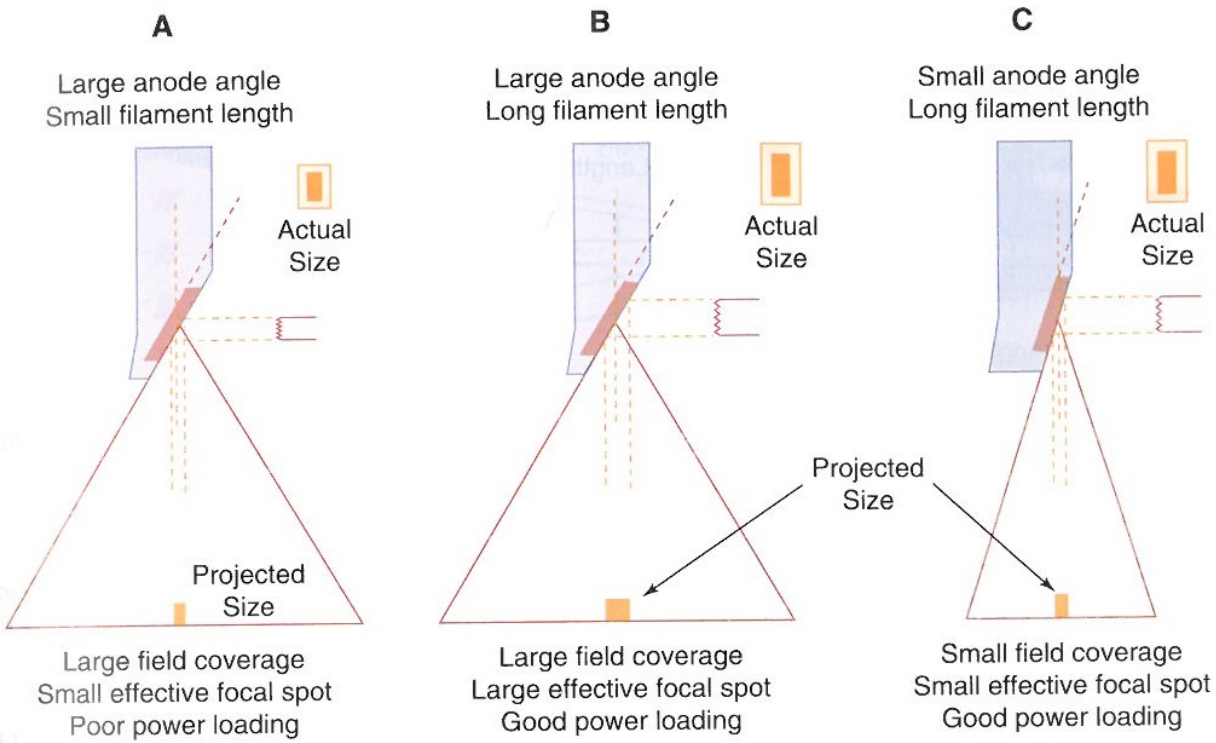



**Figure 12.8** The x-ray tube generates x rays that are restricted by the aperture in the collimator. The Al filter removes low-energy x rays that would not penetrate the body. Scattered secondary radiation is trapped by the grid, whereas primary radiation strikes the screen phosphor. The resulting light exposes the film.

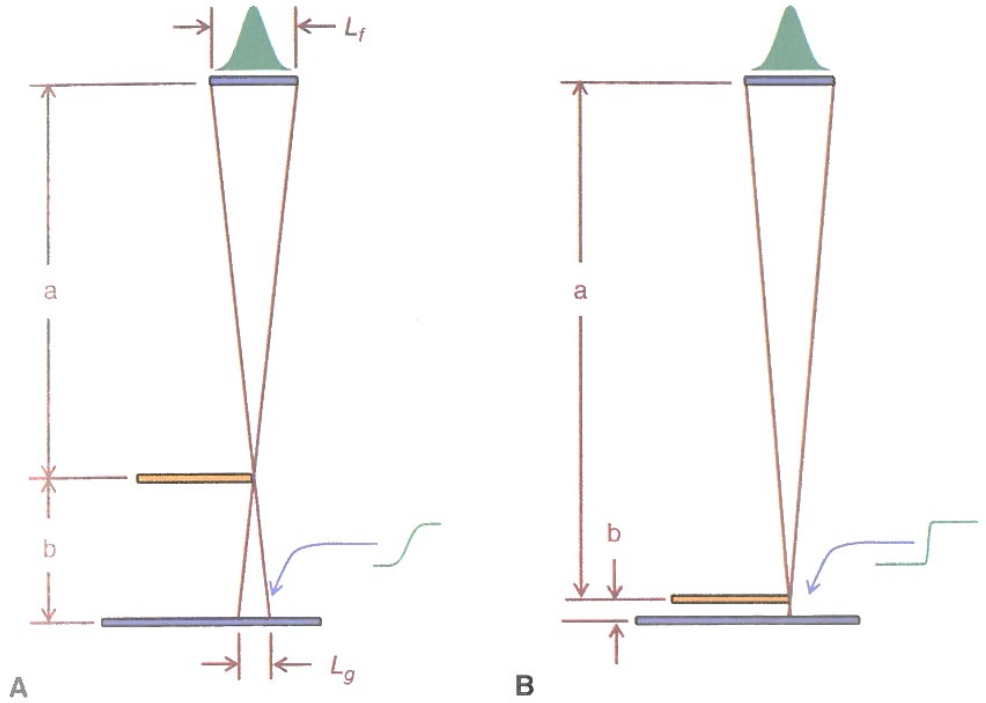
Effective focal length = actual focal length x sin $\theta$
Large focal spot allows greater x-ray beam intensity, but loss of spatial resolution that increases with the distance between object and receptor
High x-ray output allows short exposure times

# Field coverage and focal spot



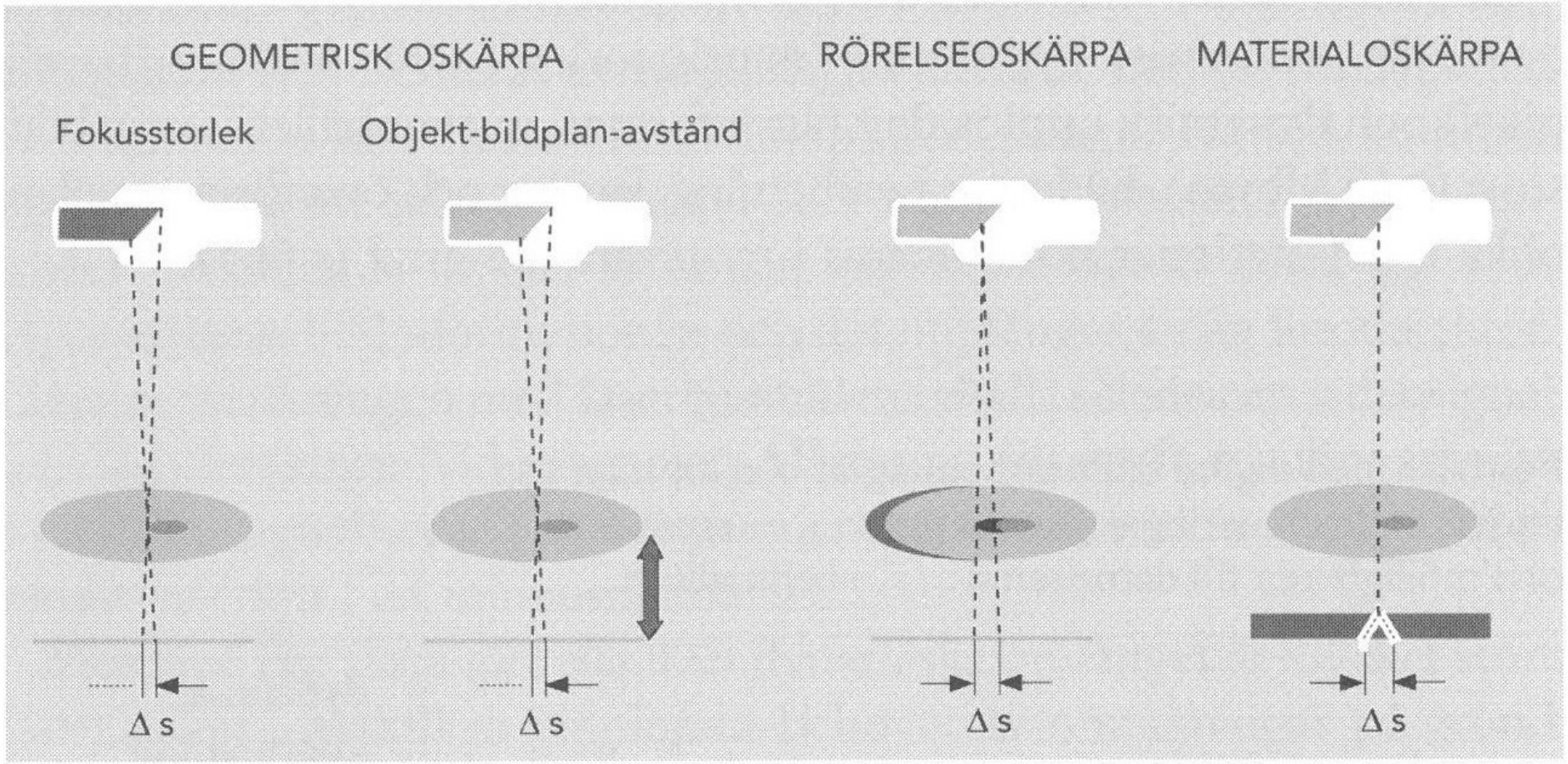

■ **FIGURE 6-14** Field coverage and effective focal spot length vary with the anode angle. **A**. A large anode angle provides good field coverage at a given distance; however, to achieve a small effective focal spot, a small actual focal area limits power loading. **B**. A large anode angle provides good field coverage, and achievement of high power loading requires a large focal area; however, geometric blurring and image degradation occur. **C.** A small anode angle limits field coverage at a given distance; however, a small effective focal spot is achieved with a large focal area for high power loading.

# Source of error: Width of focal source



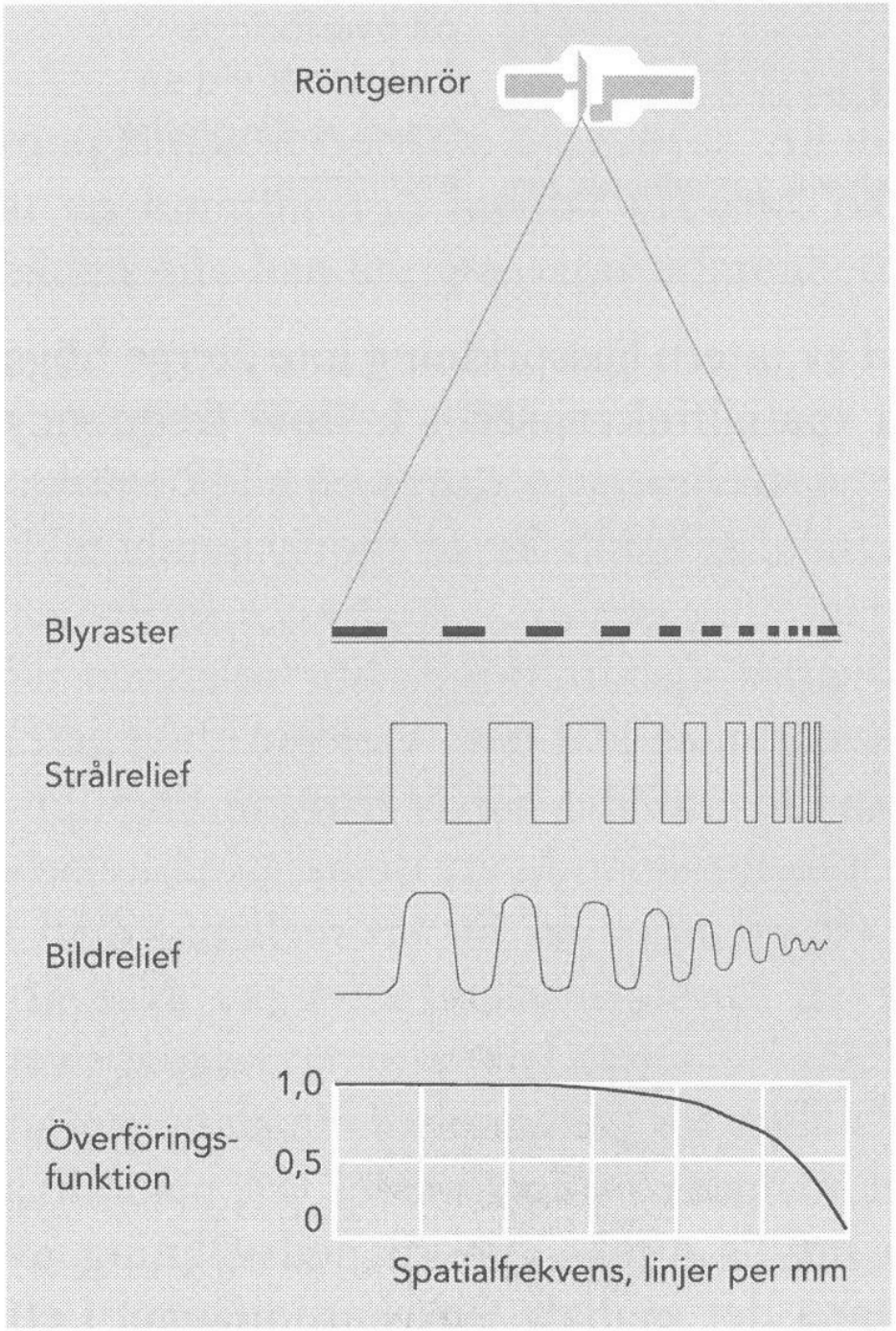

■ FIGURE 7-3 The magnification of a sharp edge in the patient will result in some blurring of that structure because of the use of a focal spot that is not truly a point source. For a focal spot with width $L_f$, the intensity distribution across this distributed (nonpoint) source will result in the edge being projected onto the image plane. The edge will no longer be perfectly sharp, but rather its shadow will reflect the source distribution. Most sources are approximately gaussian in shape, and thus the blurred profile of the edge will also appear gaussian. The length of the blur in the image is related to the width of the focal spot by $L_g = \frac{b}{a} L_f$.

# Geometrical resolution




Figur 8:32. Oskärpan i en radiologisk bild beror främst på (1) rörfokus storlek, (2) avståndet mellan objekt och bildplan, (3) rörelser hos objektet samt (4) kornstorlekar i förstärkningsskärmar och filmemulsion.

# Modulation transfer function (MTF)




Figur 8:33. Vid röntgensystem bestäms överförings-funktionen experimentellt genom att uppmäta den reproducerade kontrasten då ett linjeraster av bly avbildas.

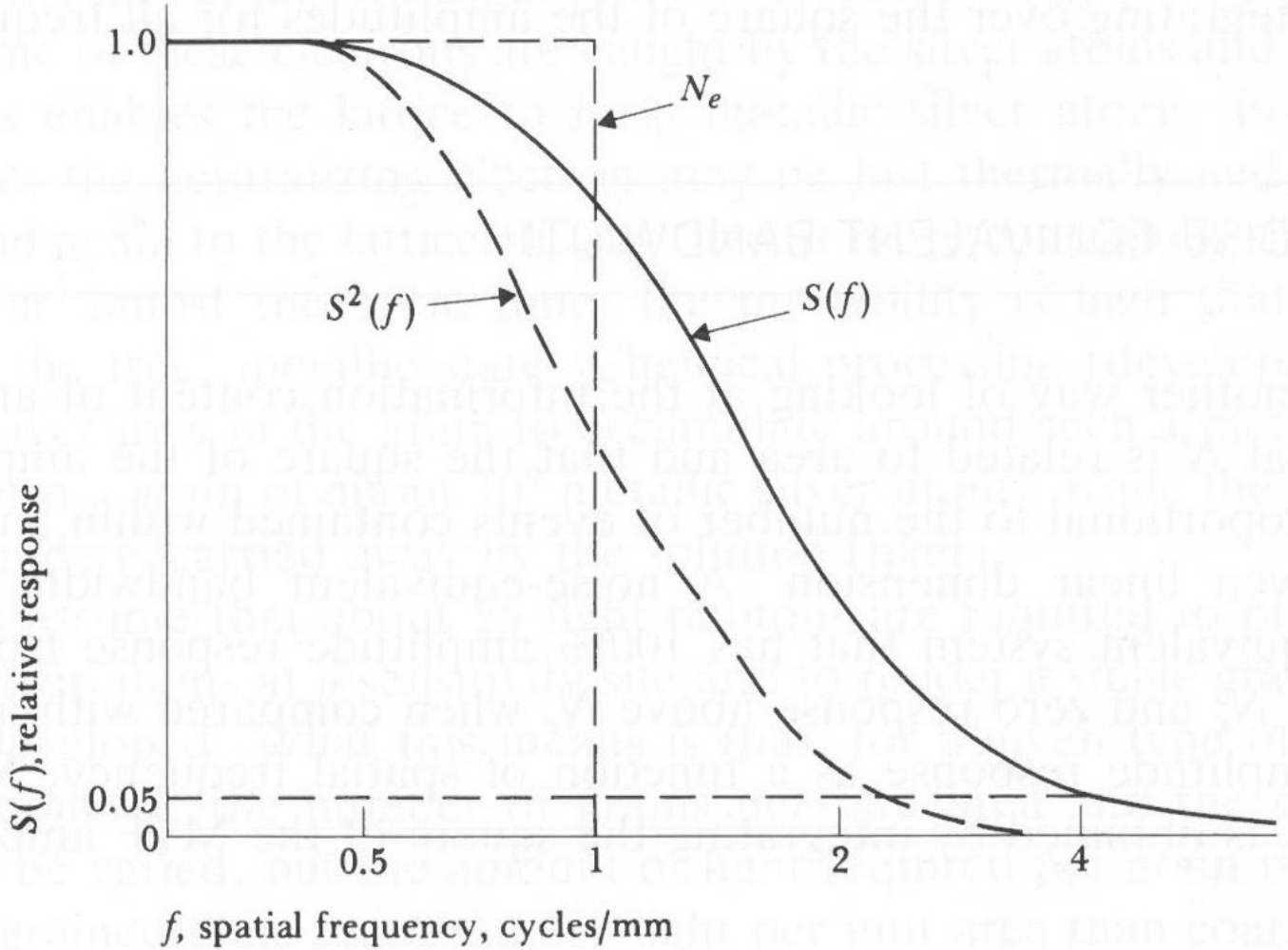
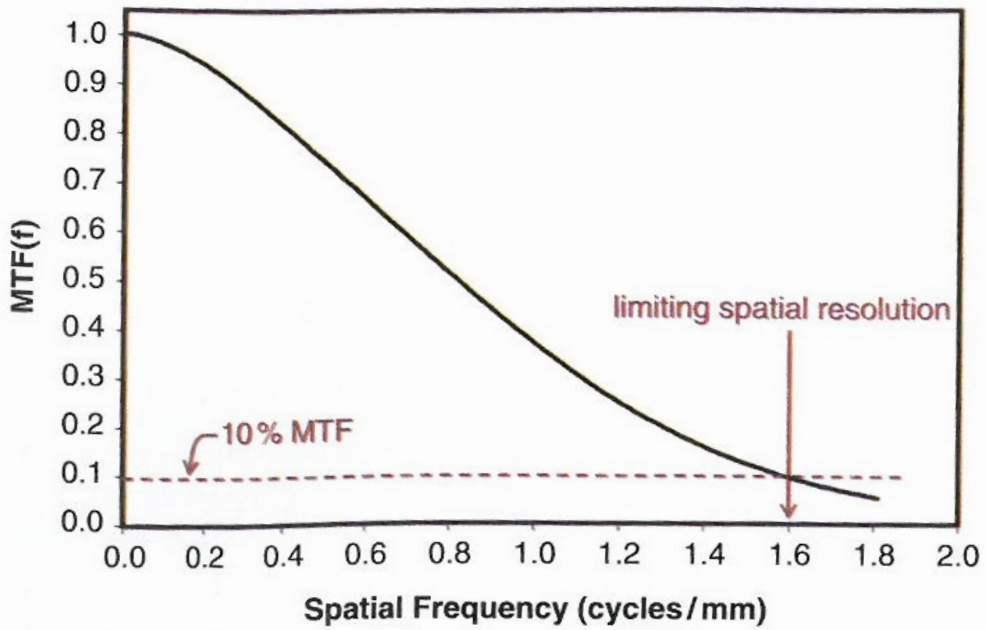



**Figure 12.4 Modulation transfer function $S(f)$ for a typical x-ray system** $S(f)$ is squared and integrated to yield $N_e$, the noise-equivalent bandwidth. The limiting resolution, 4 cycles/mm, is indicated at the 0.05 contrast level. The abscissa is plotted in cycles per millimeter, which is the same as line pairs per millimeter.

# Limiting spatial resolution






■ **FIGURE 4-14** The limiting spatial resolution is the spatial frequency at which the amplitude of the MTF decreases to some agreed-upon level. Here the limiting spatial resolution is shown at 10% modulation, and the limiting spatial resolution is 1.6 cycles/mm.

# Resolution

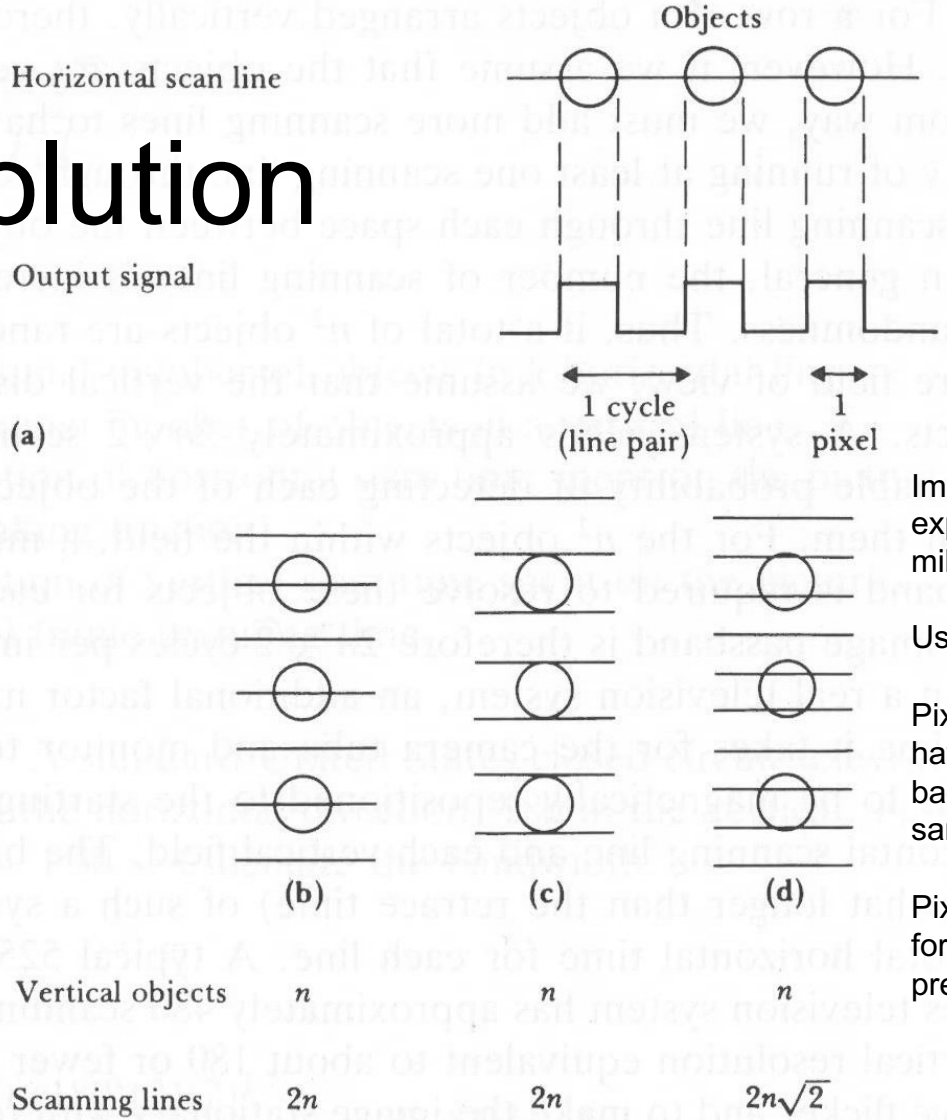


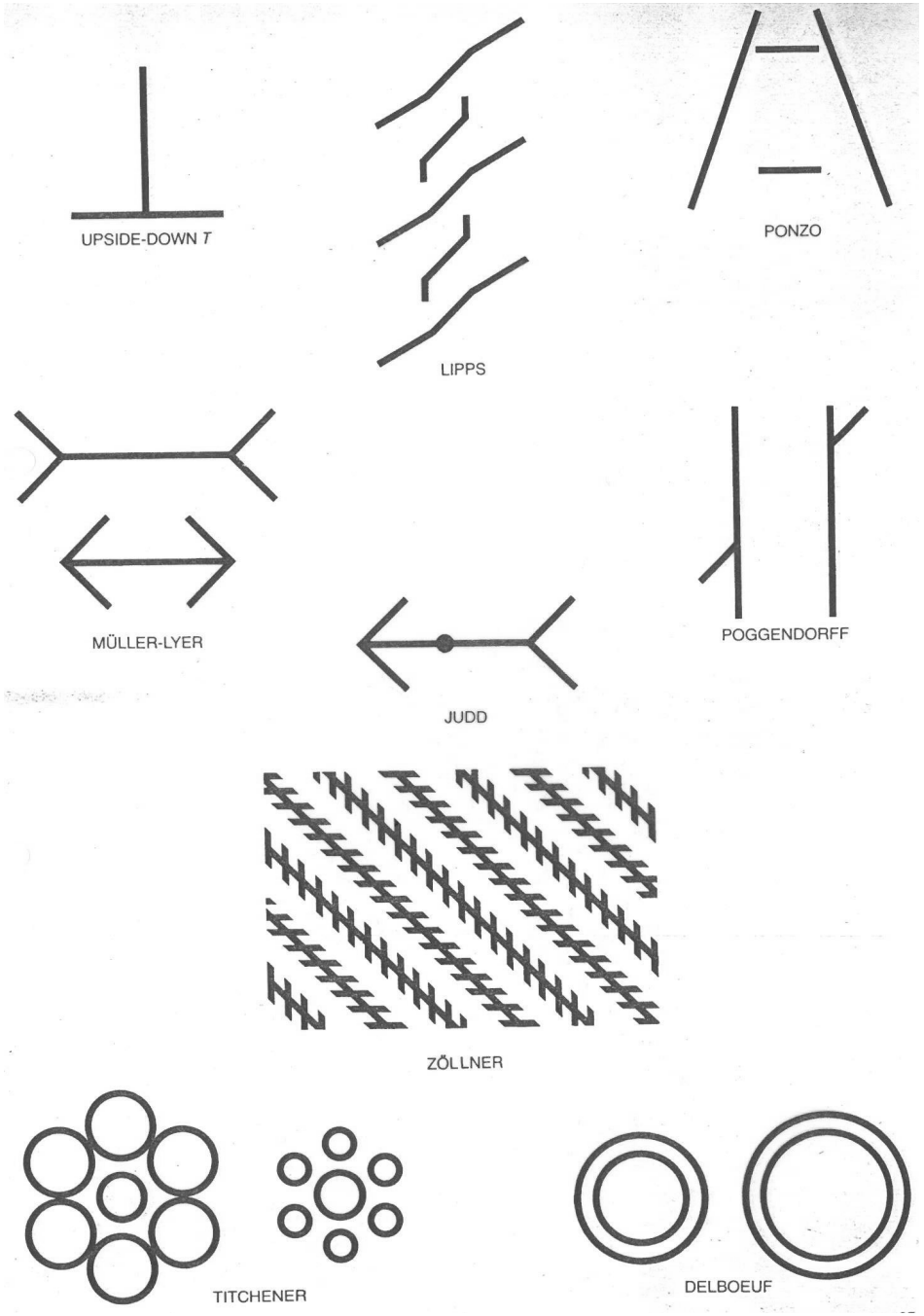

Image = surface with a spatial resolution expressed in terms of line pairs per millimileter

Using a scanner

Pixel = a space on the image surface having a dimension of one-half cycle of bandwith on the horizontal axis and the same dimension on the vertical axis.

Pixel defines the smallest size of an object for which amplitude information can be preserved

**Figure 12.1 Scanning lines and round objects** (a) Each object represents 1 pixel, but each cycle of output signal represents 2 pixels. (b) For $2n$ scanning lines, $n$ vertical objects are required. (c) If objects are located between scanning lines, $2n$ lines are insufficient. (d) For adequate resolution, $2n\sqrt{2}$ lines are required.




Cognitive prosesses and image evaluation

Objectivity?

# Contrast and noise

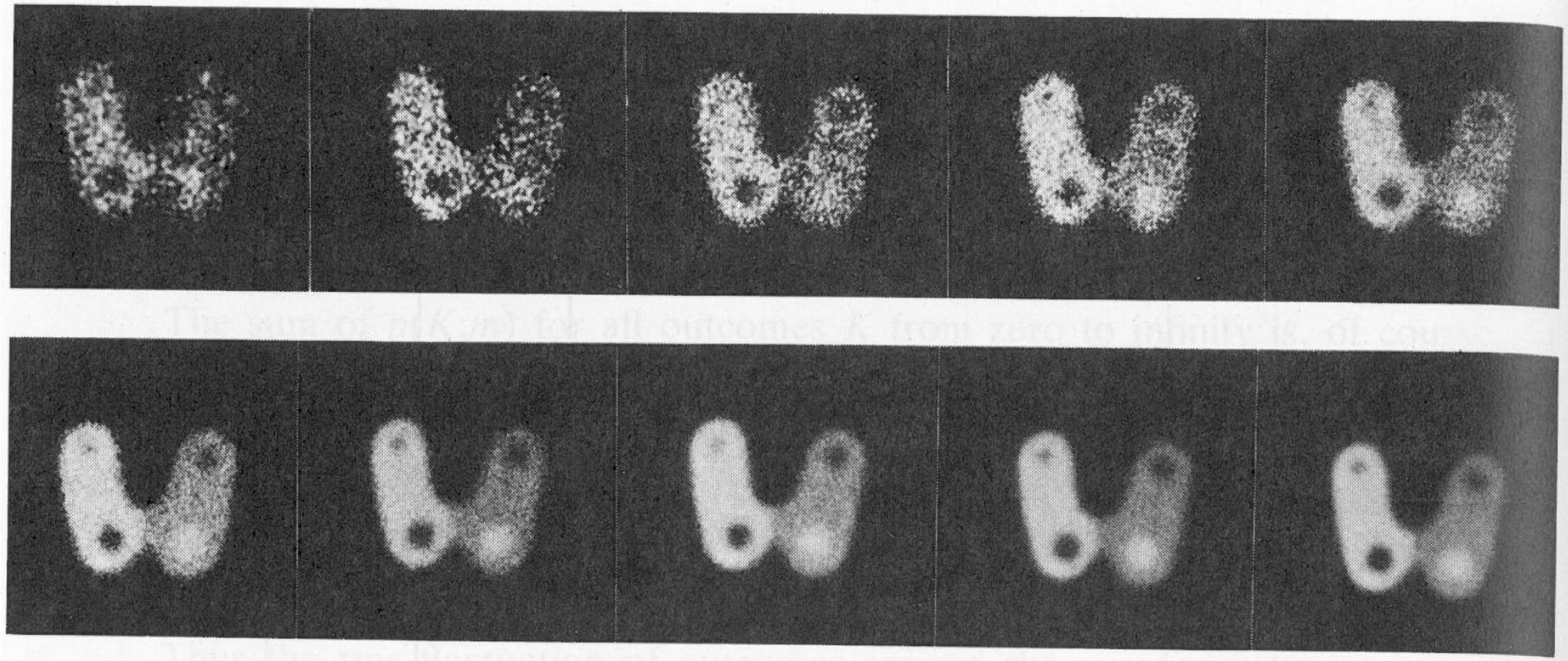
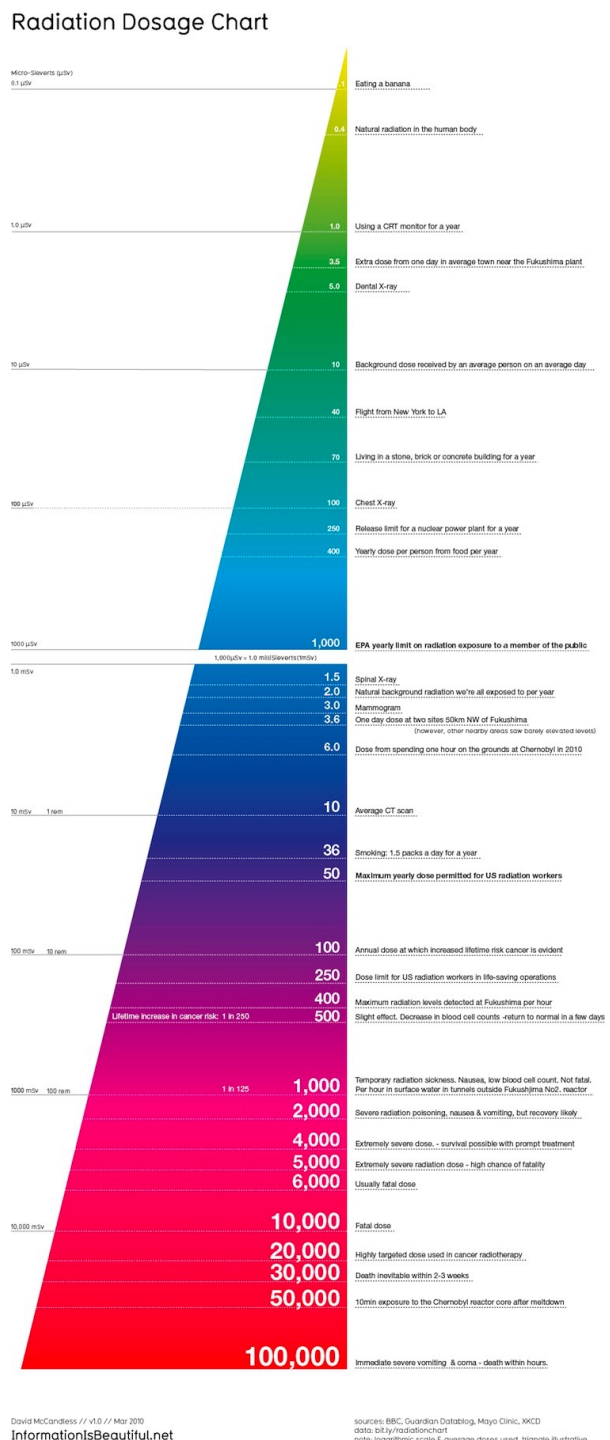
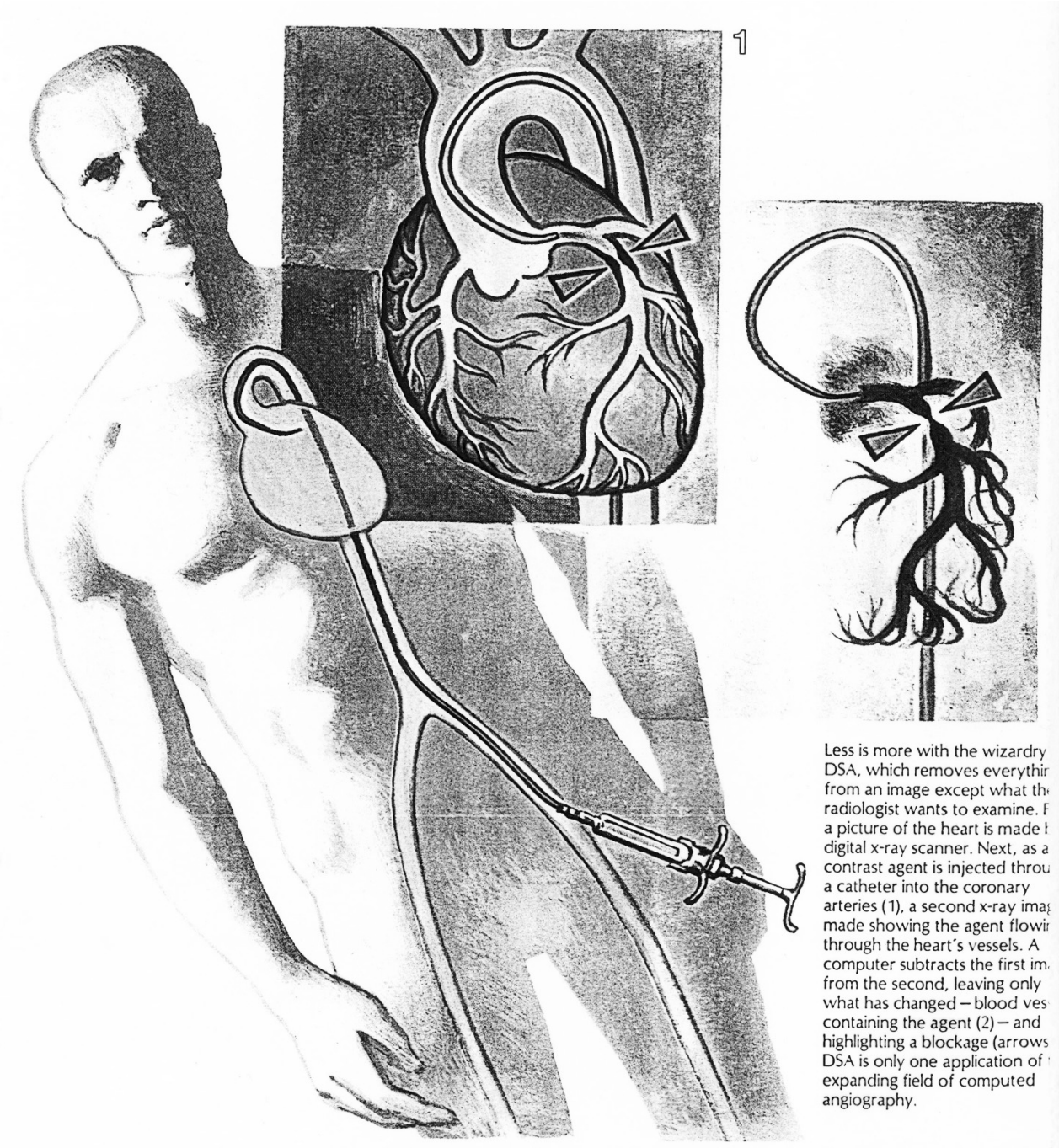


**Figure 12.3** In each successive gamma-camera picture of a thyroid phantom, the number of counts is increased by a factor of 2. The number of counts ranges from 1563 to 800,000. The Polaroid camera aperture was reduced to avoid overexposure as the number of counts was increased.




# Lethal dose?

Source: Mayo clinic

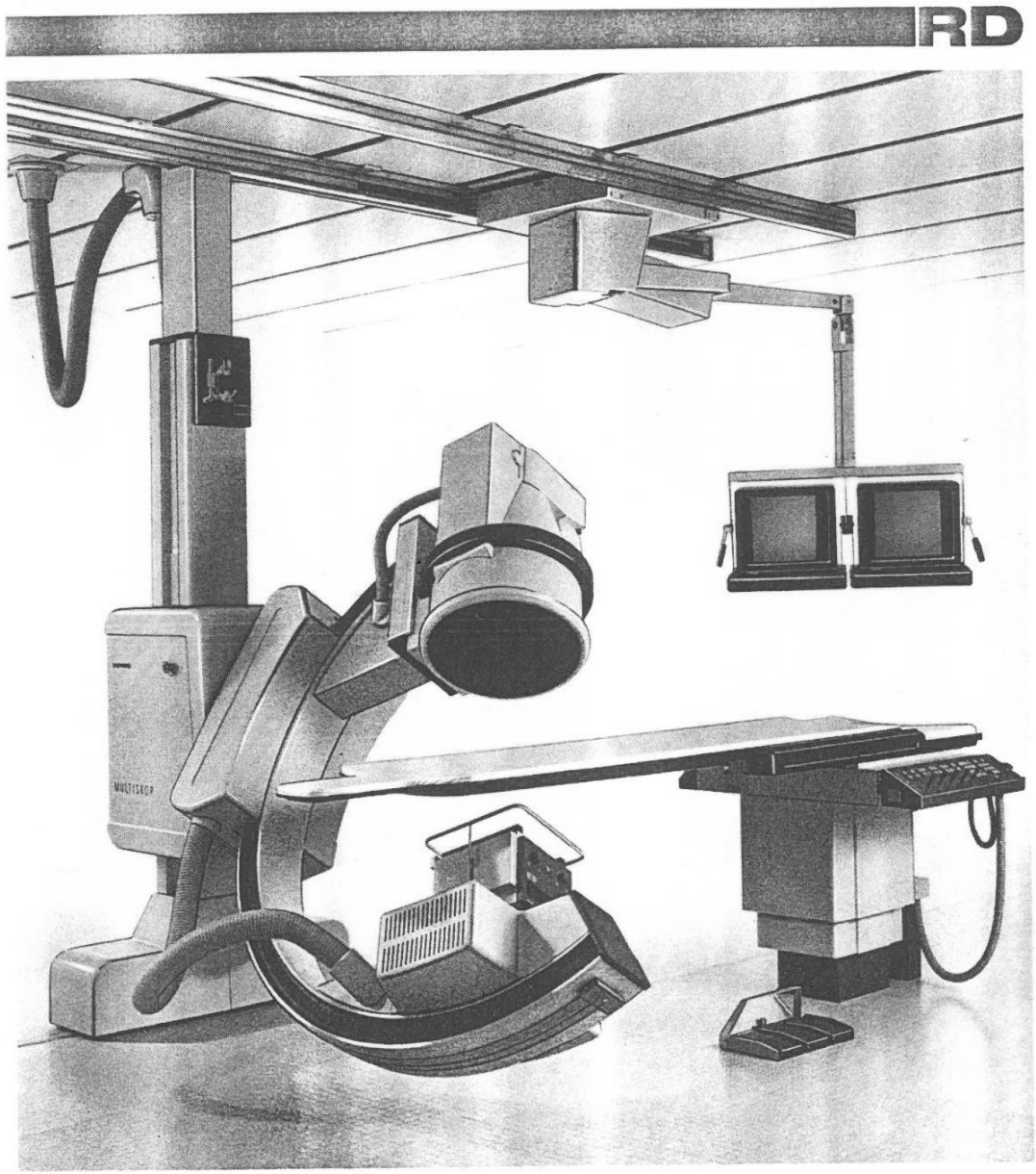



Less is more with the wizardry
DSA, which removes everythir
from an image except what th
radiologist wants to examine. F
a picture of the heart is made l
digital x-ray scanner. Next, as a
contrast agent is injected throu
a catheter into the coronary
arteries (1), a second x-ray ima
made showing the agent flowi
through the heart's vessels. A
computer subtracts the first im
from the second, leaving only
what has changed – blood ves
containing the agent (2) – and
highlighting a blockage (arrows
DSA is only one application of
expanding field of computed
angiography.

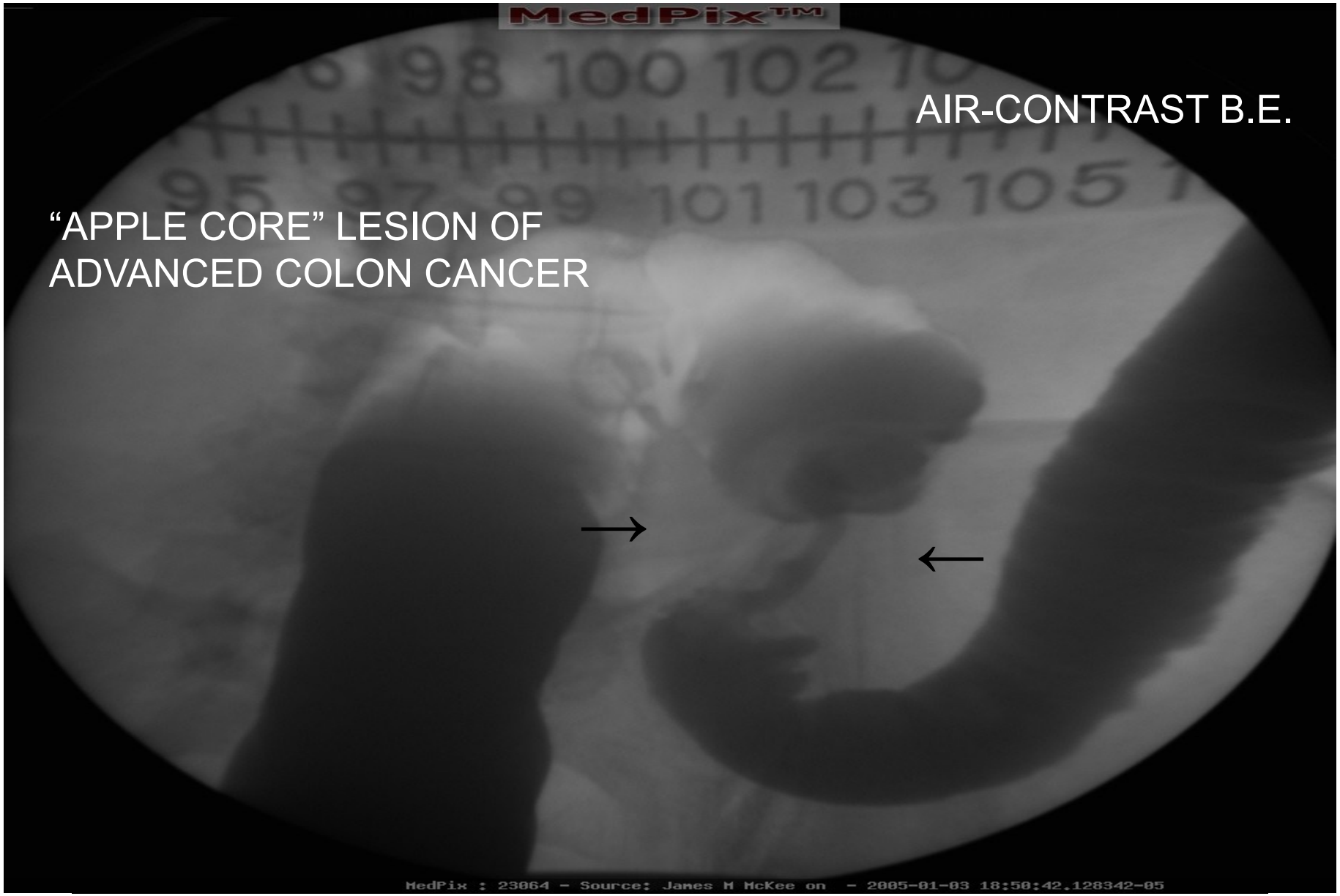
# Angio, contrast

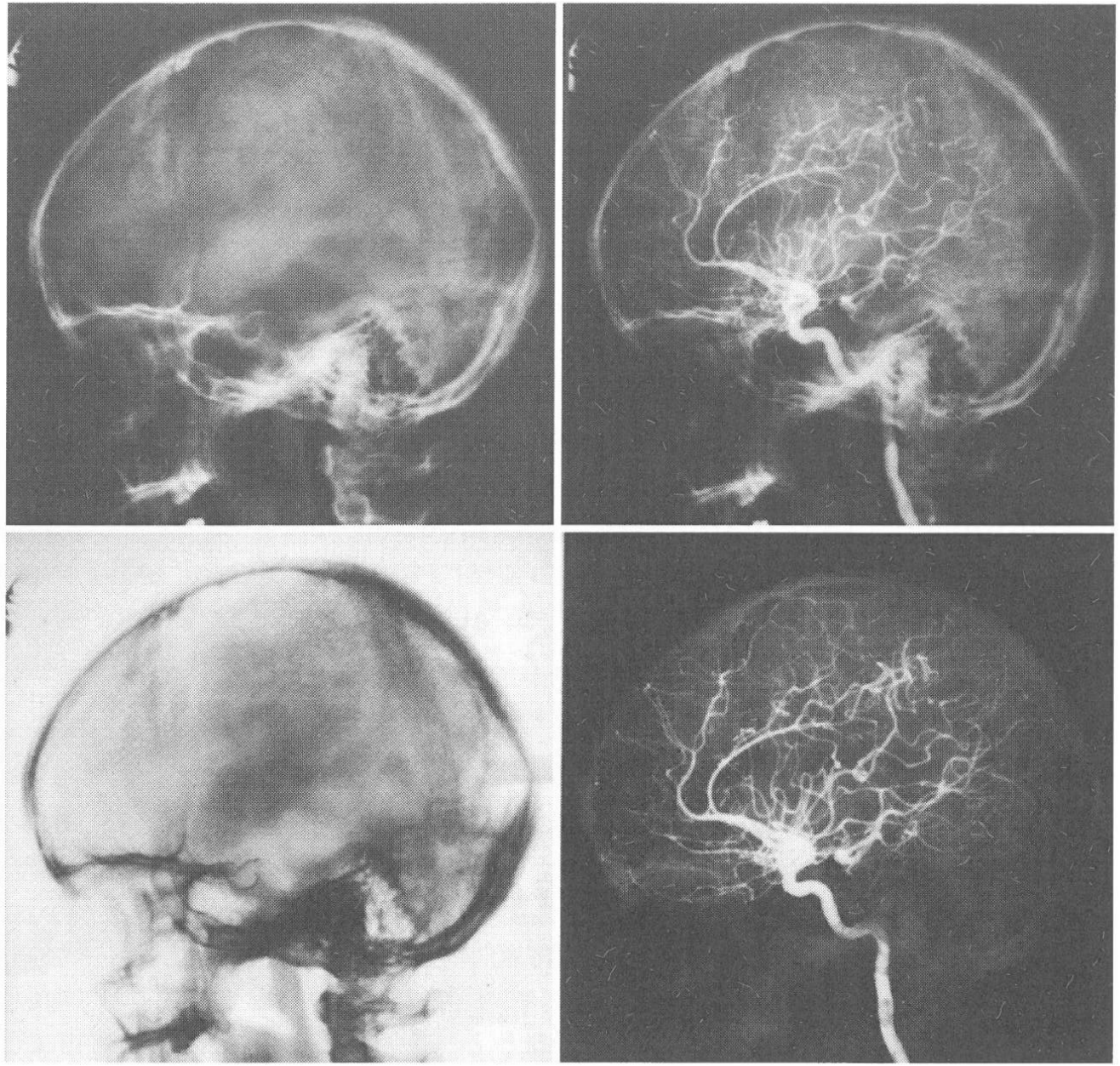
# MULTISKOP with KOORDINAT S

Angiographic/Radiologic Imaging System for Diagnostic and Therapeutic Procedures




C-arm


MedPix™
AIR-CONTRAST B.E.
“APPLE CORE” LESION OF
ADVANCED COLON CANCER
MedPix : 23064 - Source: James M McKee on - 2005-01-03 18:50:42.128342-05

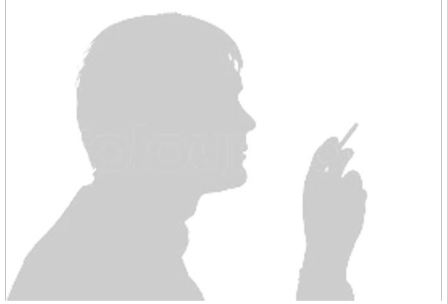


Figur 8:41. Nederst till höger angiografisk subtraktionsbild. Överst ursprungliga bilder och nederst till vänster omvändningsbilden utnyttjad vid subtraktionen.

# Subtraction image

Two pictures, one with inverted contrast is subtracted, and voila.

# Case 6



- We need more information, including images from the brain.

**What to do now?**

# Computertomograph (CT)



# What is the CT?

- Mathematical transform to the measured data.
- Reconstruct $n$ dimension function (image)
  => projection data of $n$ – 1 dimension
- Radon Transform (1917)
  “Two dimension and three dimension objects can be reconstructed from the infinite set of projection data”.
- The First CT: 1973 in the U.S.
  4 minutes scan, thickness of 10mm

# The CT-scan



# Rapidscanner

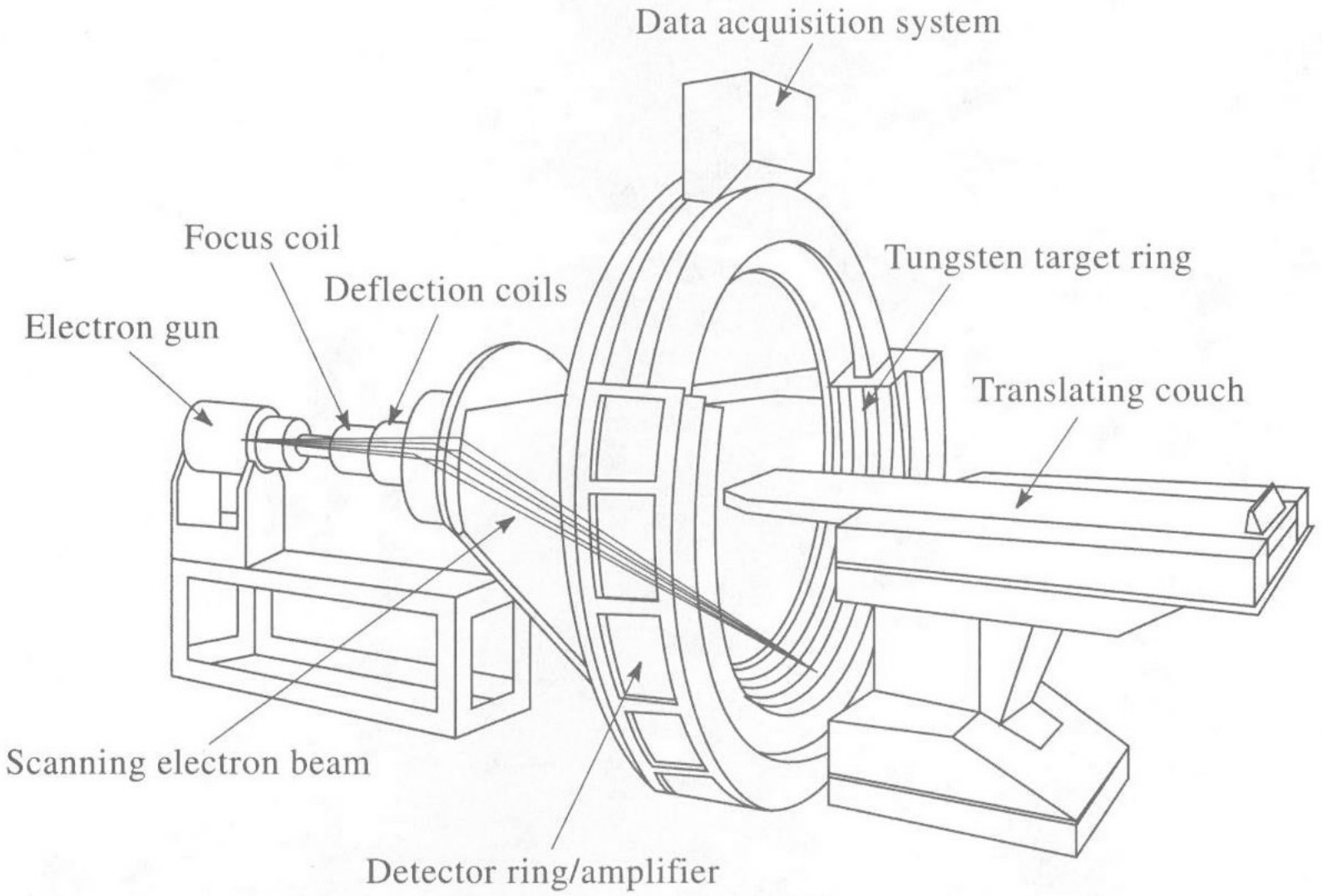
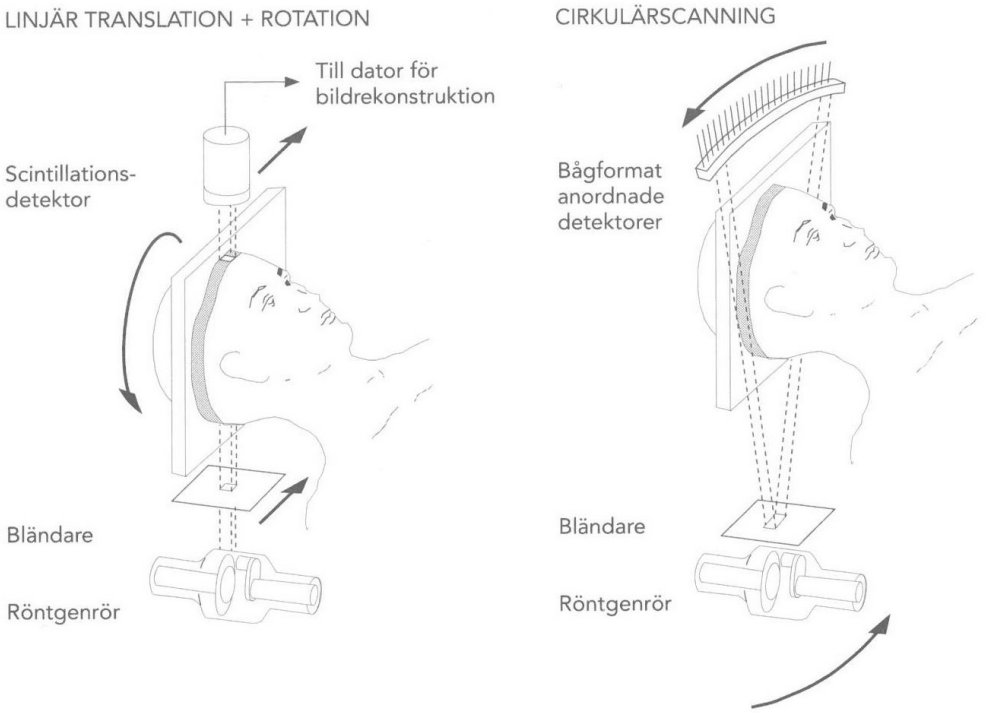


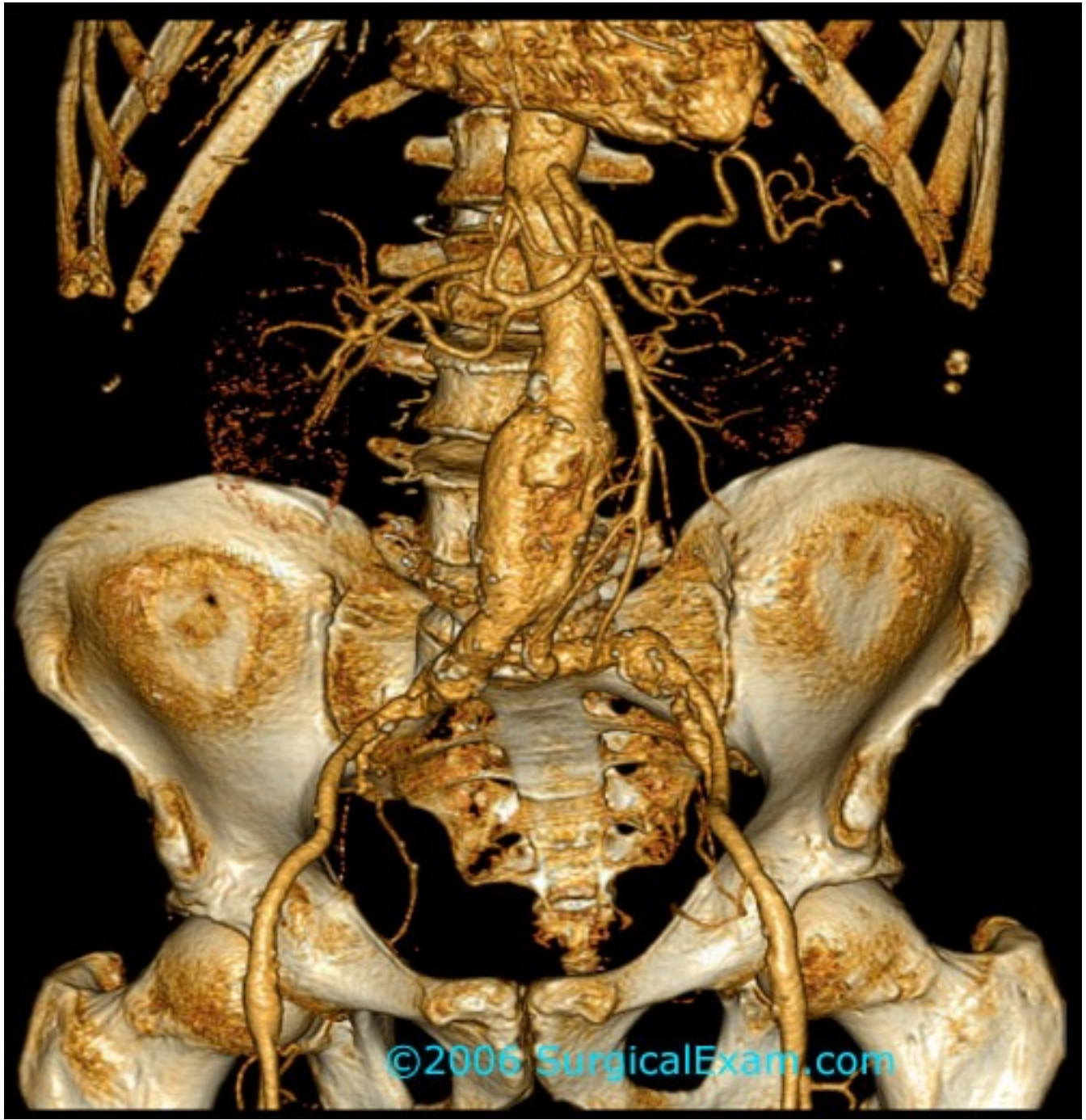

**Figure 12.13** IMATRON electron beam CT system. (Courtesy of Doug Boyd, IMATRON Corp.)




Figur 8:49. Överst till vänster den ursprungliga principen för upptagning av projektioner vid datortomografi. Överst till höger det praktiska utförandet för att öka snabbheten genom att utnyttja många bågformigt anordnade detektorer. Nederst modern CT-apparatur för volymscanning med en spiraltomograf, där två lägen visas av rör/detektorsystemets spiralrörelse.

# 1., 2. and 3. generation scanner

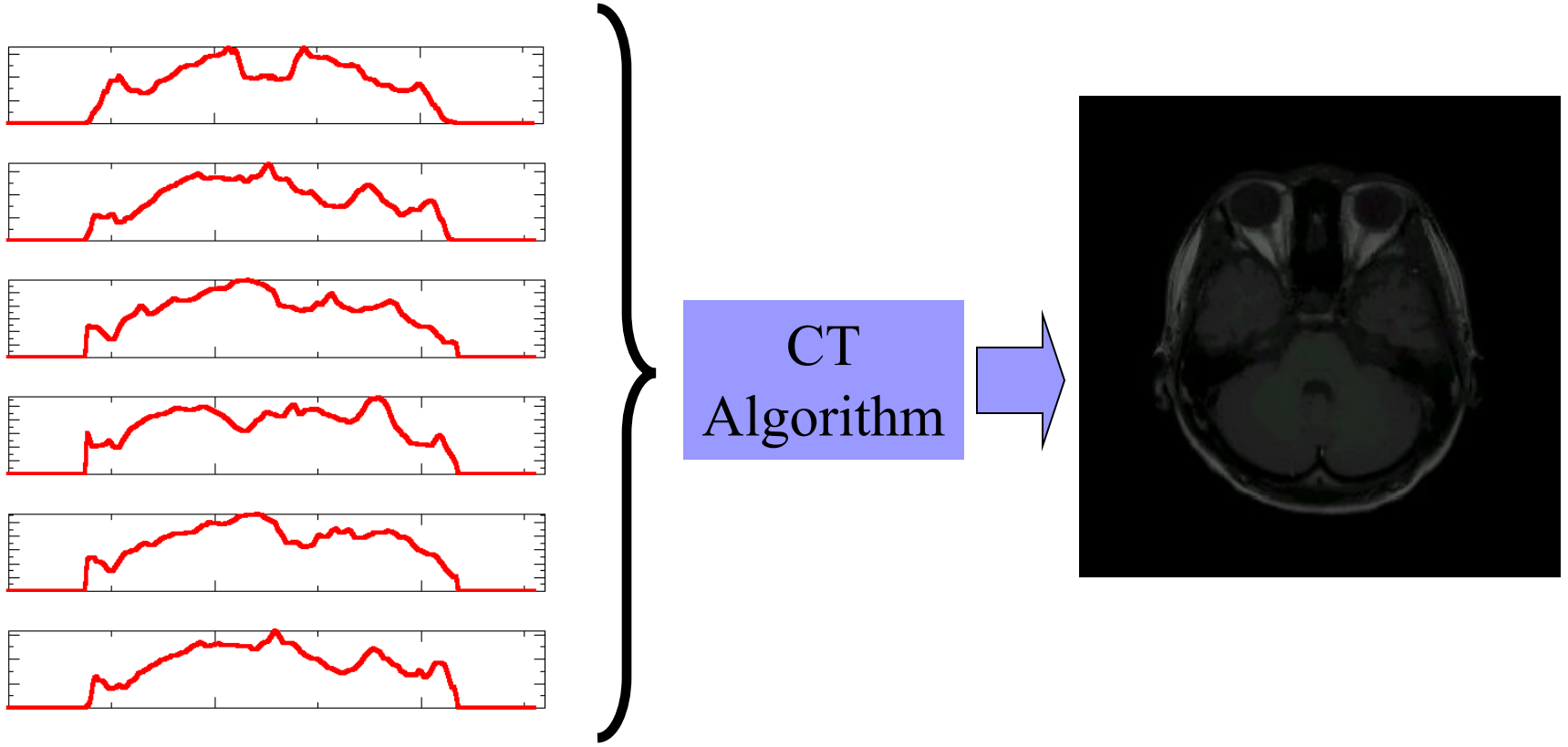
Linear translation + rotation
Circular scanning
Volume scanning




CT
3-D
image

# Concept of CT

- Getting the shape by back projection of  the projection data.
- For example, outward view is the quadrangle => it is the cylinder



# Geometry of a CT-scanner

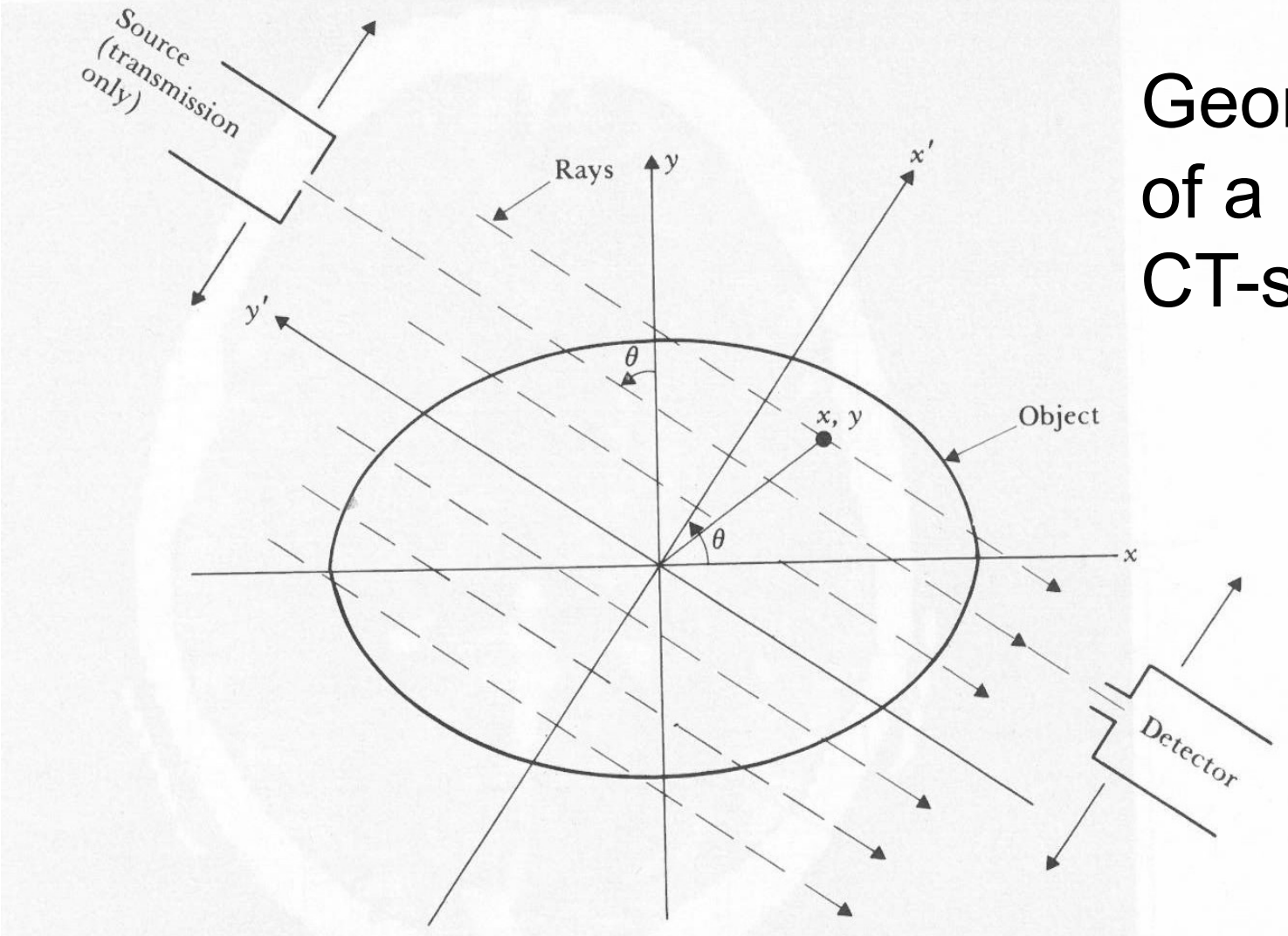


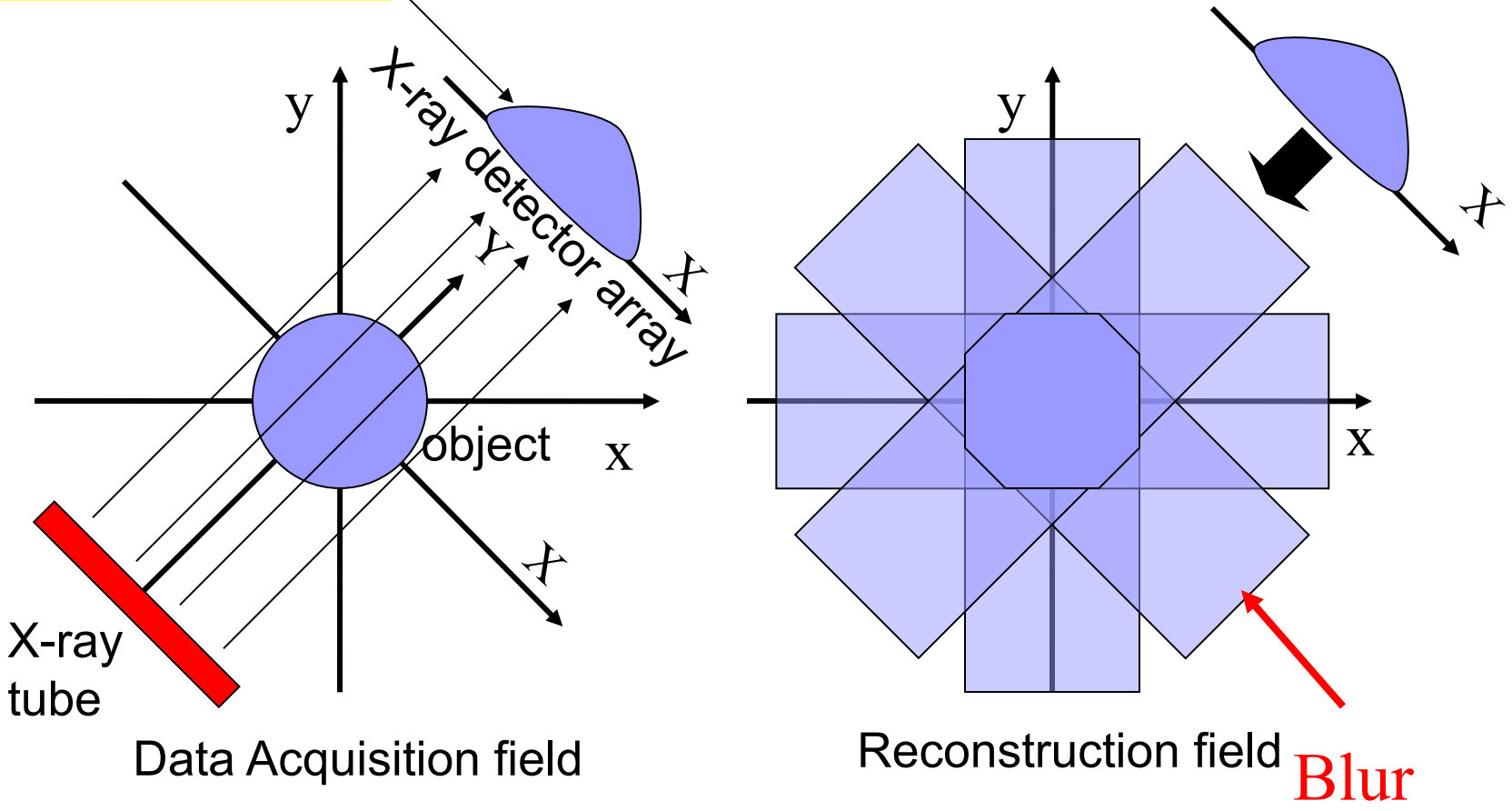

**Figure 12.13 Basic coordinates and geometry for computed tomography** The projection rays shown represent those measured at some angle $\theta$. The source and detector pair are rotated together through a small angle, and a new set of rays measured. The process is repeated through a total angle of 180°. (From R. A. Brooks and G. Di Chiro, "Theory of image reconstruction in computed tomography," *Radiology*, 1975, 117, 561–572.

# Basic principle of CT

-Reconstruction of 2 dimensional image-

Projection Data

curvilinear integral of absorption coefficient regarding Y




Simple Backprojection

# Back-projection (1)

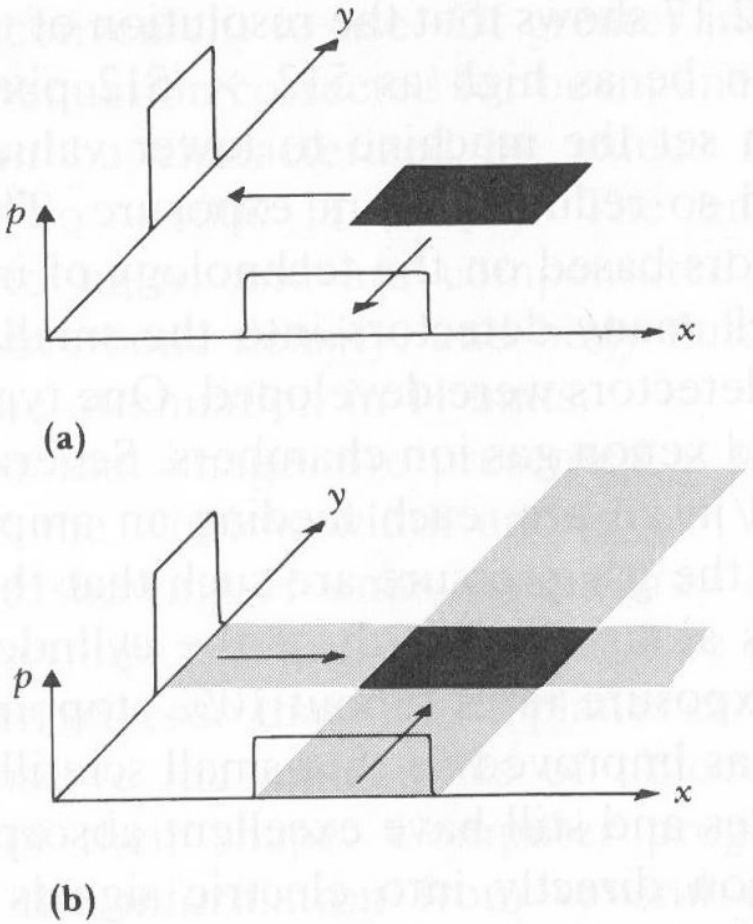
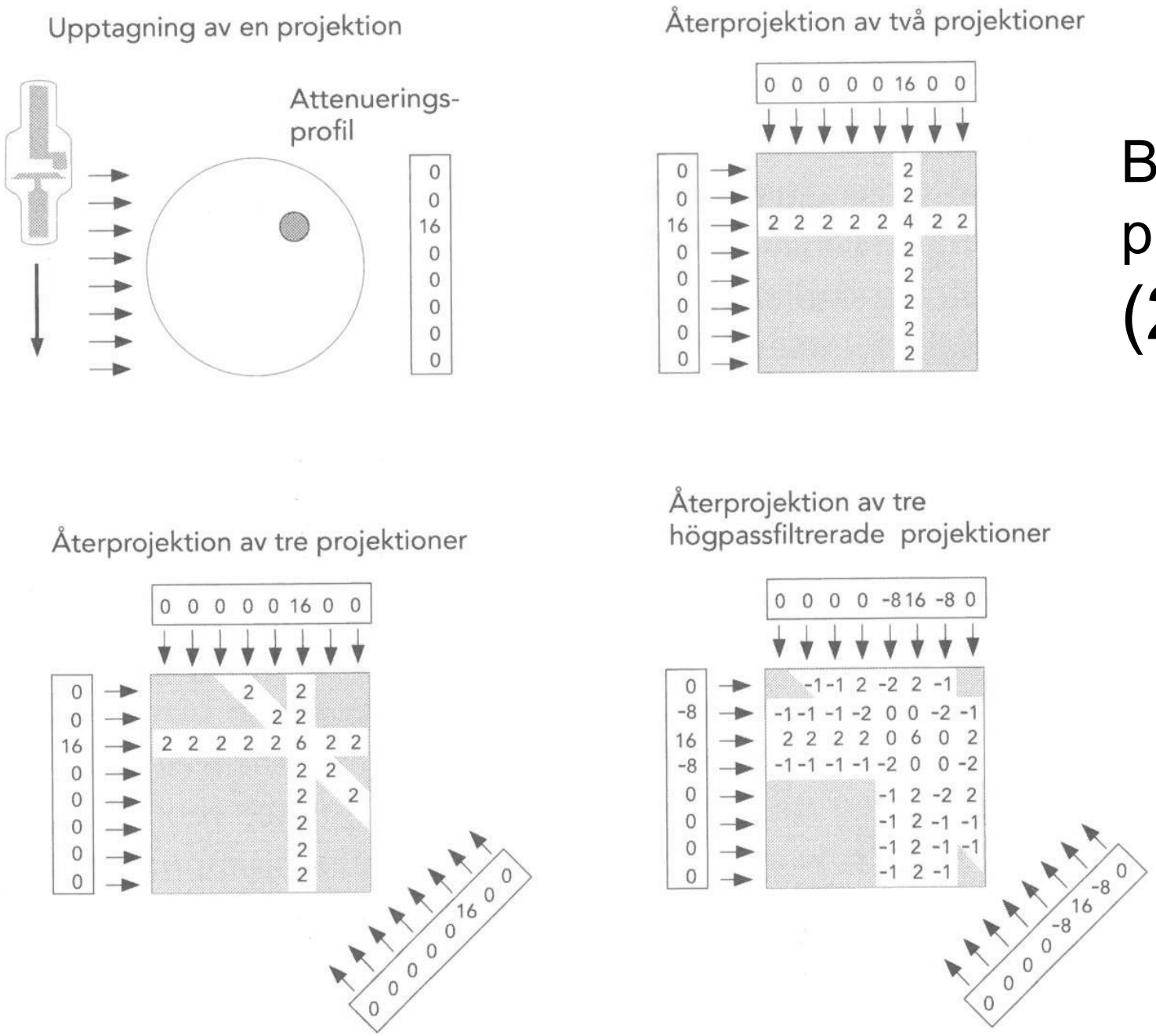



**Figure 12.16 Back projection** (a) Projections of this object in the two di tions normal to the $x$ and $y$ axes are measured. (b) These projection data projected back into the image plane. The area of intersection receives t summed intensities. It is apparent that the back-projected distribution is alre a crude representation of the imaged object. (From R. A. Brooks G. Di Chiro, "Theory of image reconstruction in computed tomography," *diology*, 117, 1975, 561–572.)

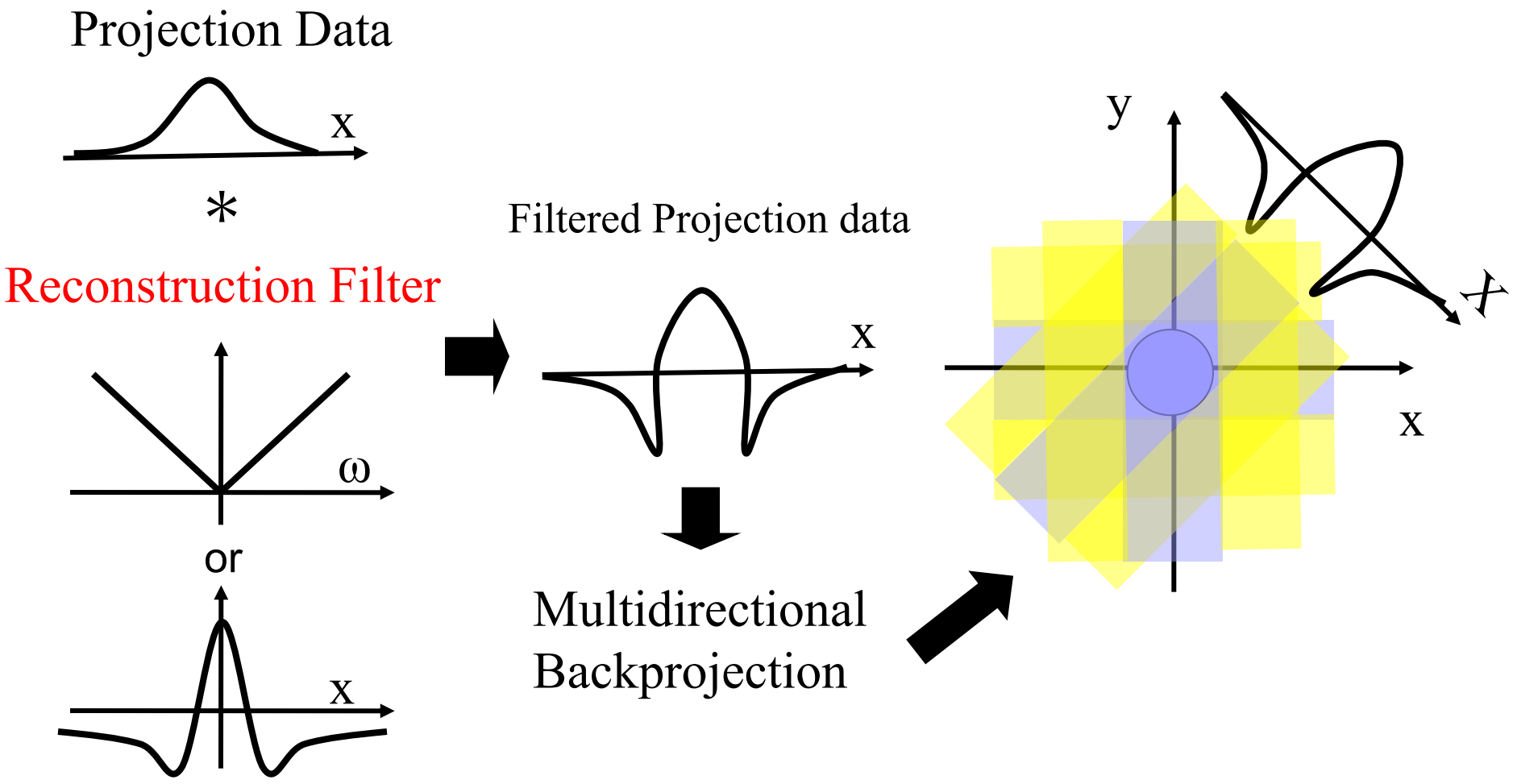



Back-projection (2)

Figur 8:50. Rekonstruktion av CT-bild utan och med filtrering (P Edholm).

# Basic principle of CT

-Reconstruction of 2 dimensional image-

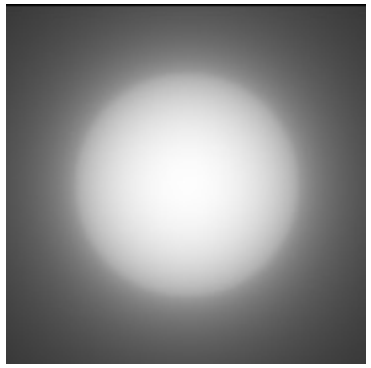



Filtered Backprojection

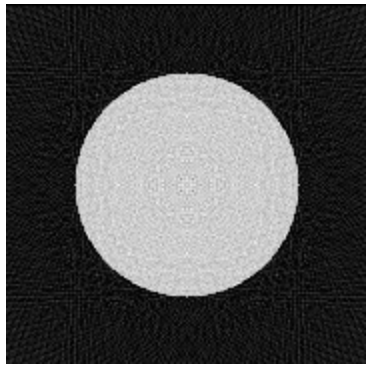
# Example of Simulation



Model Image

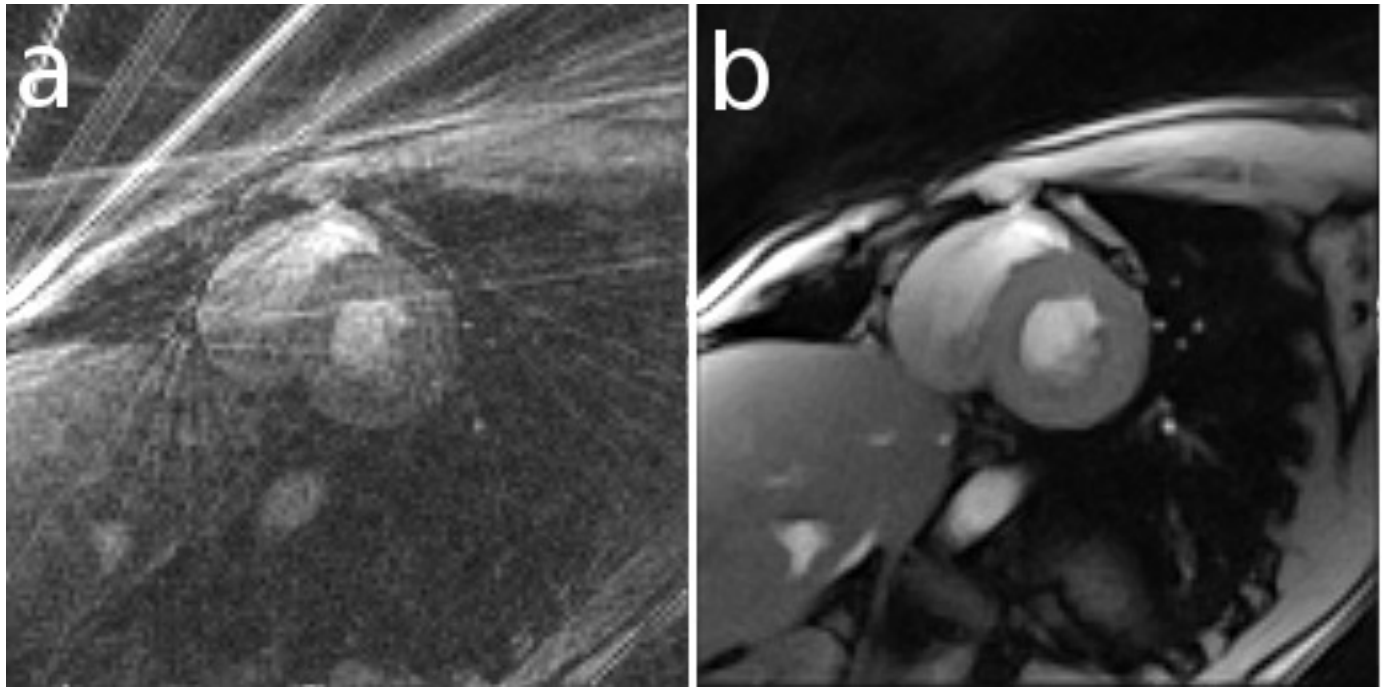


Simple Backprojection



Filtered Backprojection

# Iterative techniques - CT




https://en.wikipedia.org/wiki/Iterative_reconstruction#/media/File:Heart-direct-vs-iterative-reconstruction.png

**Figure 2**

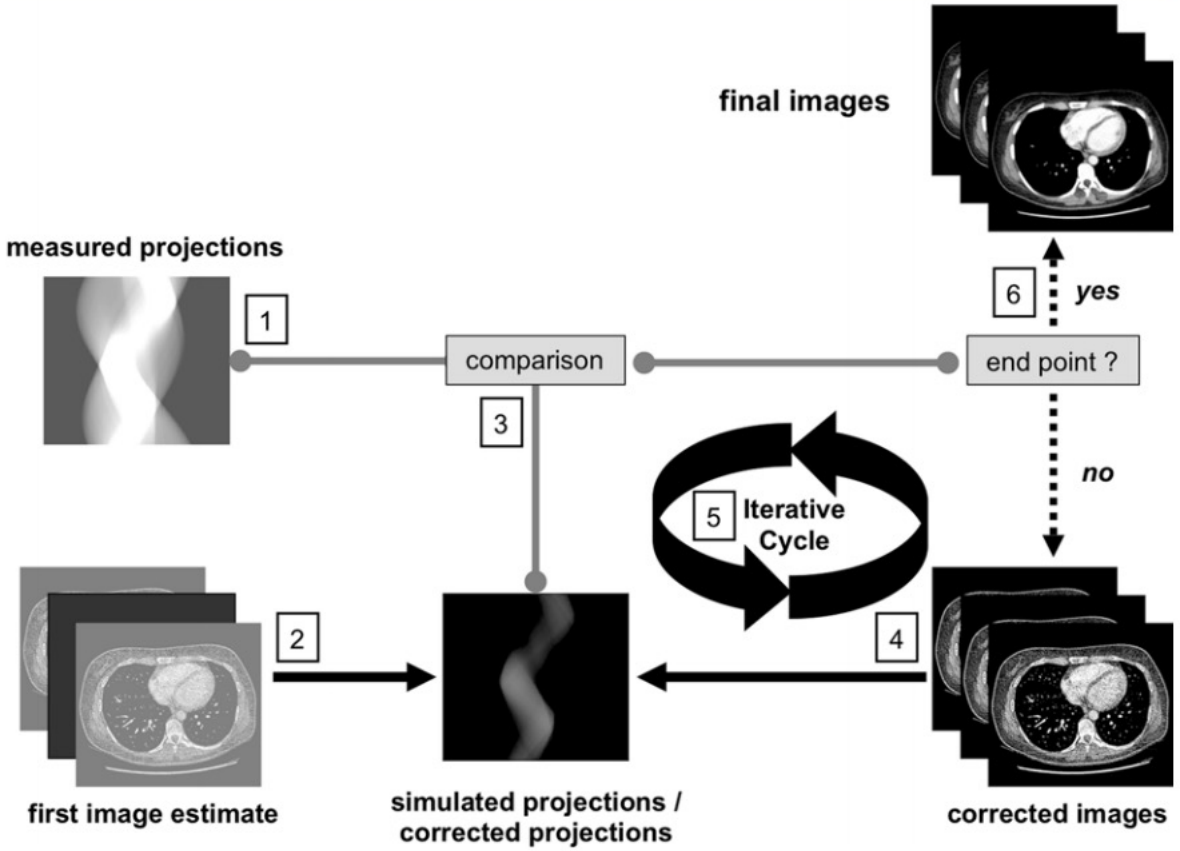



**Figure 2:** Schematic representation of the principle steps of iterative image algorithms. Following the CT acquisition process (measured projections), a first image estimate is generated. An x-ray beam is simulated via forward projection to obtain simulated projection data, which are then compared with the measured projection data. In case of discrepancy, the first image estimate is updated based on the characteristics of the underlying algorithm. This correction of image and projection data is repeated until a condition predefined by the algorithm is satisfied and the final image is generated.

Iterative = An initial guess about the two-dimensional pattern of x-ray is made. Projection data predicted by this guess are then calculated and used for comparison with the measured results. Discrepancies between the measured and the modelpredictions are employed in a continuous iterative improvement of the model array

# Iterative process

Imagine views are collected from 4 angles, 0, 45, 90 and 135. Each measurement is represented as the sum of voxel attenuations. This leads to six equations:

$X_1 = U_1 + U_3$ and so on, 6 equations with 4 unknown. Can be solved mathematically, but timeconsuming when the matrix is getting large.

The iterative algortihm (ART = Algebraic Reconstruction Technique) is demonstrated for a 2x2 matrix

The last estimate match all measurments -> accepted

What is the weakness? Due to random error, no iteration will match perfectly

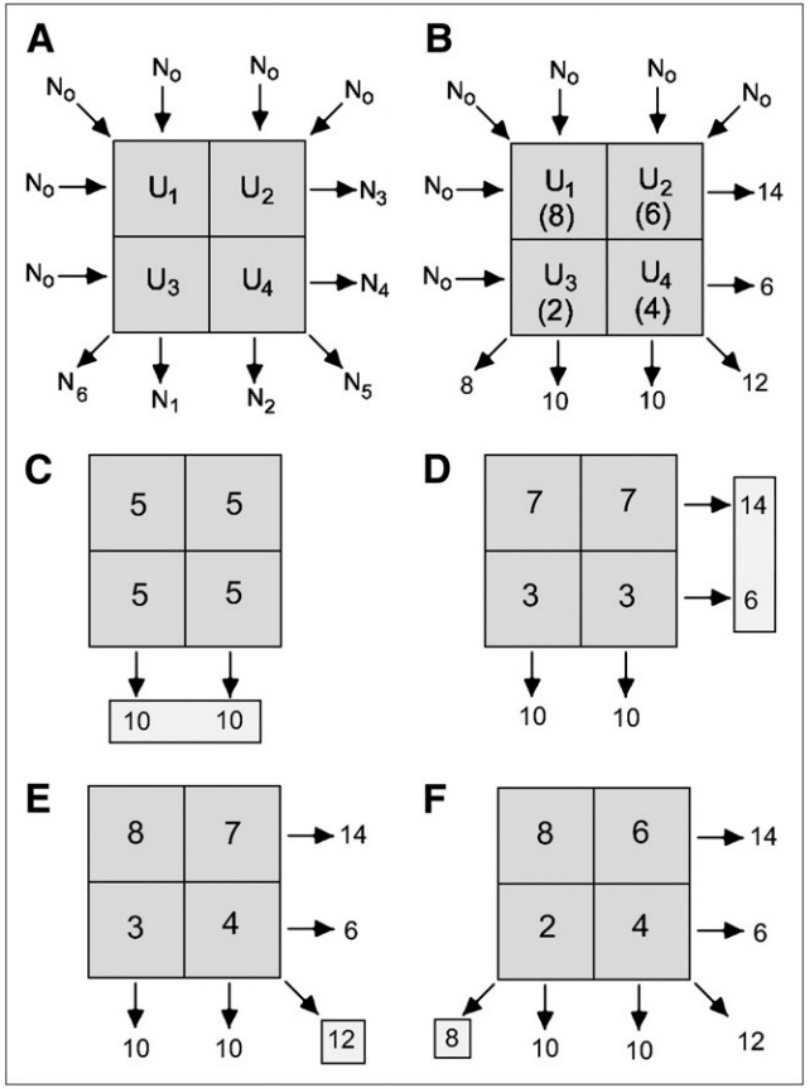
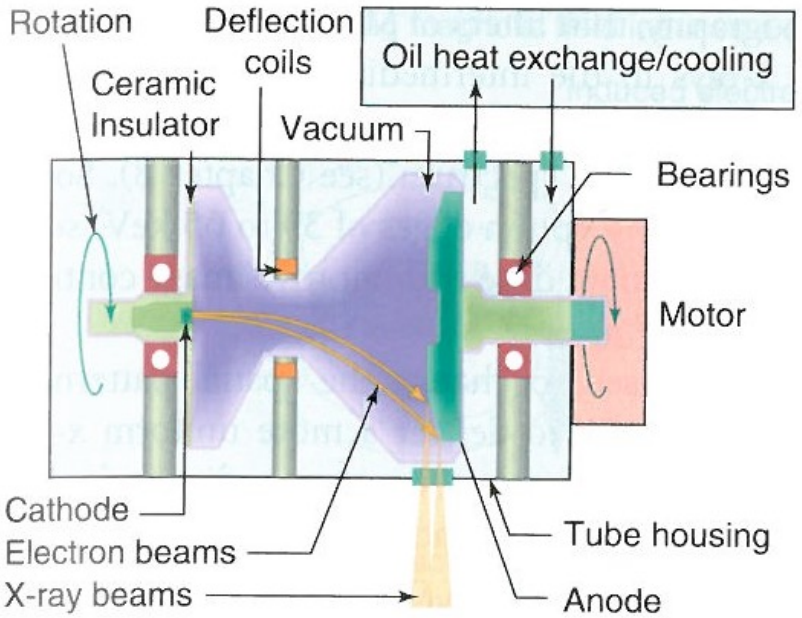



**FIGURE 4.** ART. (A) ART algorithm for 4-voxel "patient." (B) Attenuation measurements. (C) Starting estimate is constructed by dividing measurements from first view equally along their ray paths. (D–F) This estimate is iteratively adjusted to match measurements for each consecutive view, stopping when transmission measurements predicted by current estimate match all actual measurements to within some preset tolerance.

Lee Goldman. J Nucl Med Technol 2007; 35:115–128

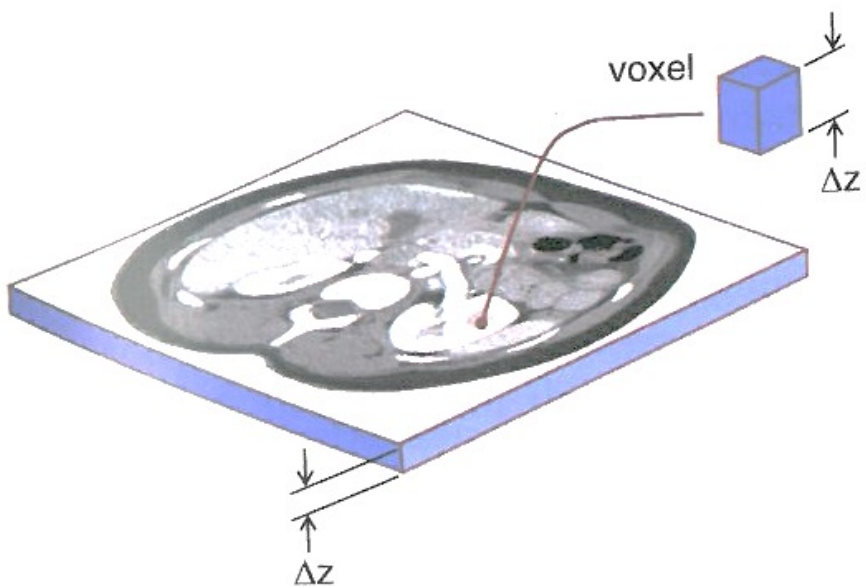
# CT x-ray tube




■ **FIGURE 6-20** Diagram of an advanced CT x-ray tube, showing the anode and the planar cathode within a rotating vacuum enclosure. The bearings are mounted outside of the vacuum enclosure. Deflection coils magnetically direct the electron beam to specific areas on the target. Circulating oil rapidly removes excess heat from the anode. The electron beam can be rapidly deflected between two focal spots; this is known as a "flying focal spot."

Bushberg et al, The essential physics of medical imaging

# Voxels

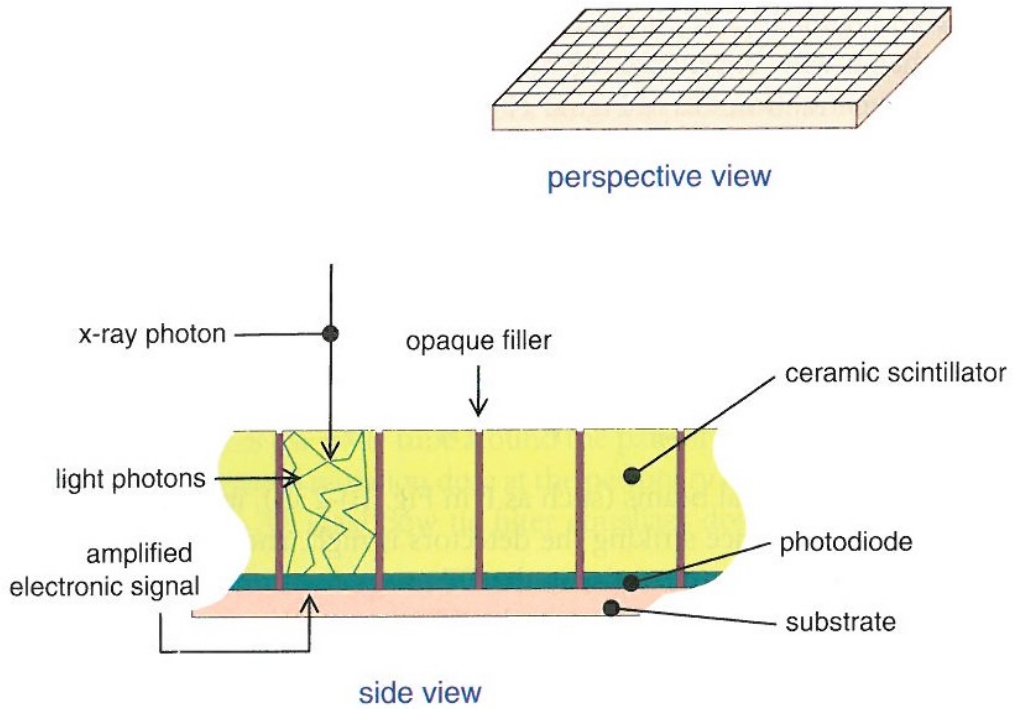



**A.** in the patient

■ **FIGURE 10-21** **A.** The dimensions of the in-plane (Δx, Δy) and the slice thickness in the z-axis (Δz) are combined to form the volume element or *voxel*. In modern MDCT scanners, the dimensions of the voxel are approximately equal on thin-slice reconstructions, where $\Delta x = \Delta y \approx \Delta z$.

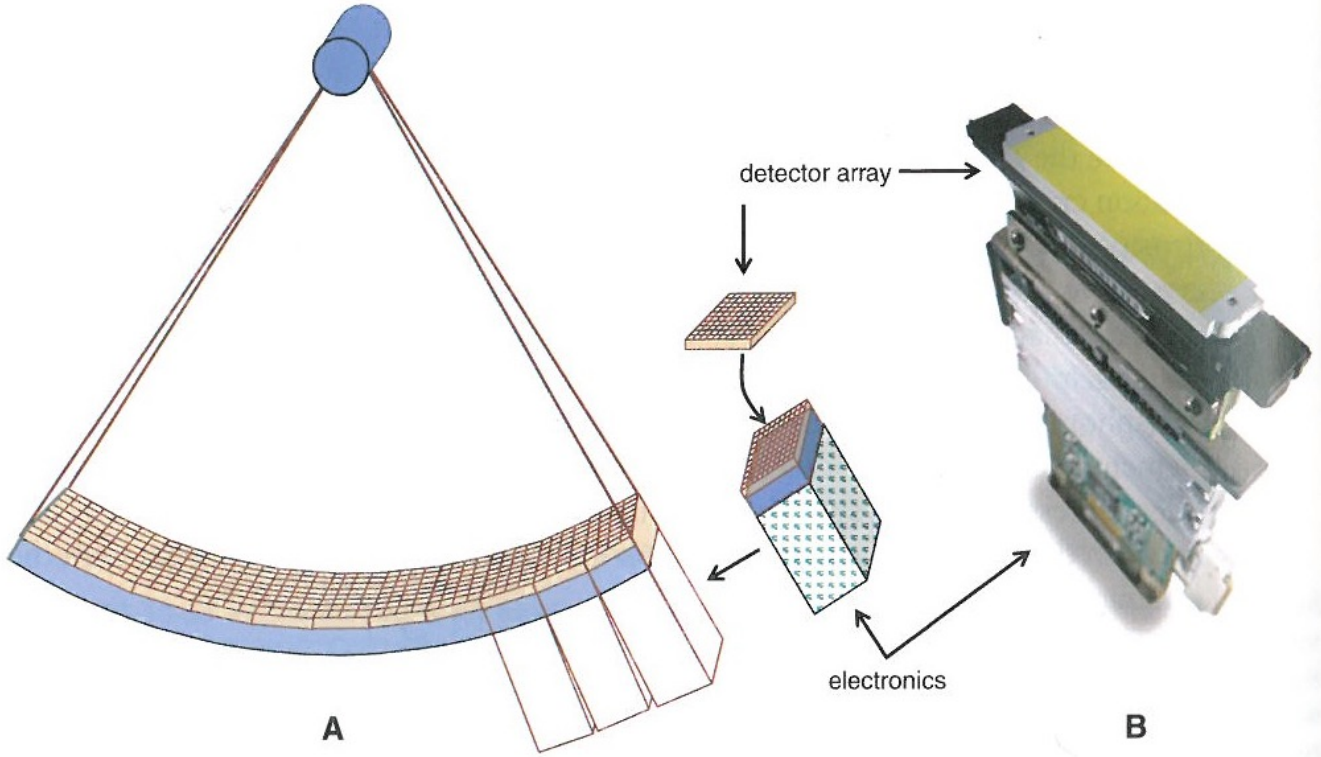
Bushberg et al, The essential physics of medical imaging

# Detector



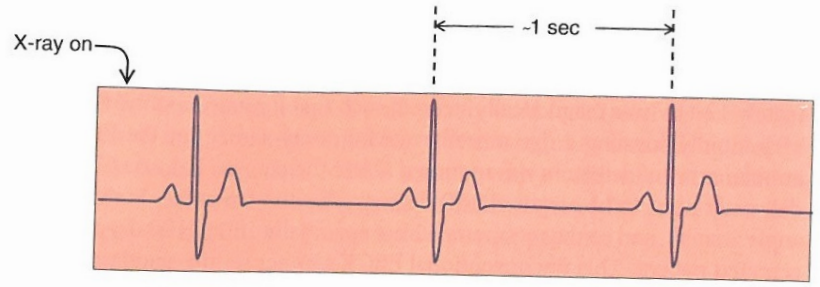

■ **FIGURE 10-24** The detectors in modern CT are fabricated as detector modules (perspective view). A side view of the modern CT detector shows individual detector elements coupled to photodiodes that are layered on substrate electronics. Spaces between detector elements are scored out creating a small void, which is filled with an optical filler material to reduce detector cross talk. CT detectors are much more expensive than those used in radiography and make use of sintered phosphors resulting in high-density ceramic detector arrays. The increased density improves x-ray detection efficiency.

# Detector array

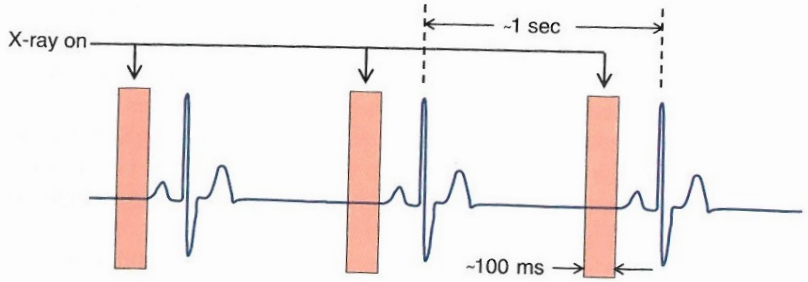



■ **FIGURE 10-25** **A.** The detector arrays are mounted on electronics module, which includes power supplies for the amplifiers, amplification circuits for the detector module, and analog-to-digital converter systems. **B.** A photograph of a detector module with electronics from a commercial CT scanner is shown. (Photograph courtesy J. Hsieh.)

# Cardiac CT



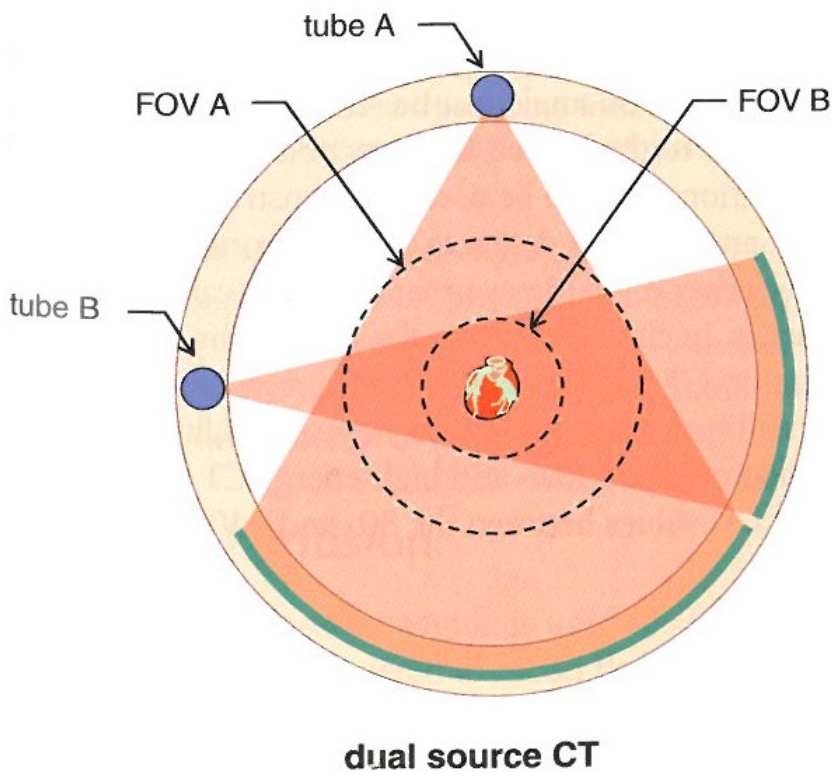

**B.** retrospective cardiac gating




**C.** prospective cardiac gating

■ **FIGURE 10-38** (*Continued*) **B.** Early cardiac CT acquisition modes used retrospective gating, where the CT scanner acquires data while it also records the electrocardiogram (ECG) information. This allowed the acquired projection data to be selected at various time points in the cardiac cycle ("gated"), and these projection data are then combined and used to reconstruct cardiac CT images—with retrospective gating, CT images could be produced throughout the cardiac cycle, resulting in the generation of a 3D movie of the beating heart. However, this acquisition mode results in high dose to the patient and there is only occasional need for cardiac imaging throughout the entire cardiac cycle. **C.** In prospective cardiac gating modes, the ECG is used to trigger the x-ray tube such that projection data are only acquired during the most quiescent part of the cardiac cycle, end-diastole. This mode reduces dose to the patient considerably and produces one cardiac CT image that is useful for assessing coronary artery stenosis and other anatomical anomalies in the heart.

# Cardiac CT




■ **FIGURE 10-39** The dual source CT system was developed specifically for cardiac imaging. This system has two full x-ray imaging systems—two x-ray tubes and two sets of detector arrays; however, due to space issues, the B imaging chain has a smaller detector array resulting in a smaller field of view, but still well tailored to the imaging requirements for the heart. Due to the approximately 90-degree juxtaposition of the two imaging chains, a 90-degree rotation of the gantry results in the acquisition of projection data over 180 degrees, which is sufficient for true CT reconstruction at the center of the field of view. With a 330-ms period for 360-degree gantry rotation, the 90-degree rotation requires 330/4 = 83 ms. Thus, it is a short enough time period to acquire a full projection data for CT reconstruction within a fraction of the typical cardiac cycle.

# Multislice CT

- Several rings of detectors for multiple scanning of several layers.
- Time efficiant, a 4 slice multislice CT is up to 8 times faster than a single slice CT
- Possible monitoring of a beating heart ++

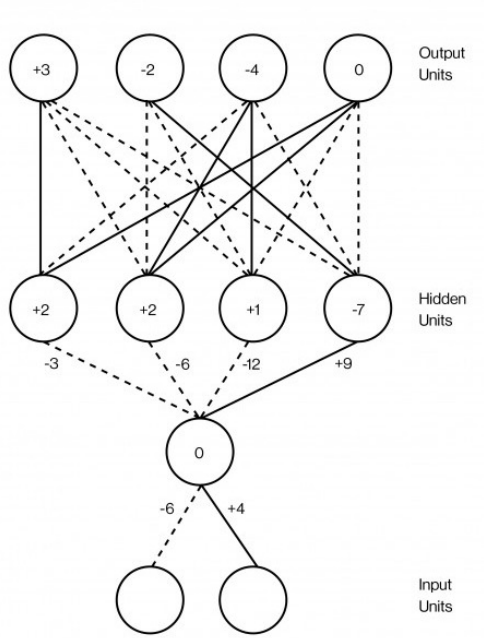


# Artificial intelligence in radiology



# Artificial intelligence in radiology

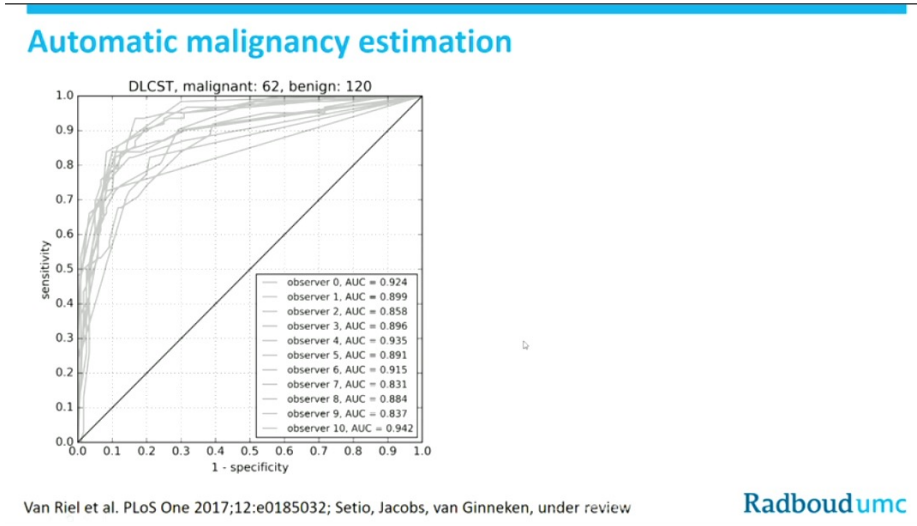





Geoffrey Hinton, chief scientific advisor University of Toronto, and research team of Google Brain

# Example of AI

Radiologi som eksempel:







Reciever operating curves (ROC) – mål på ytelse for et klassifiseringssystem

# Example of AI

Radiologi som eksempel:

