Medical Image Analysis (MIA)

https://svn.tugraz.at/svn/MedicalImageAnalysisVOKU2016

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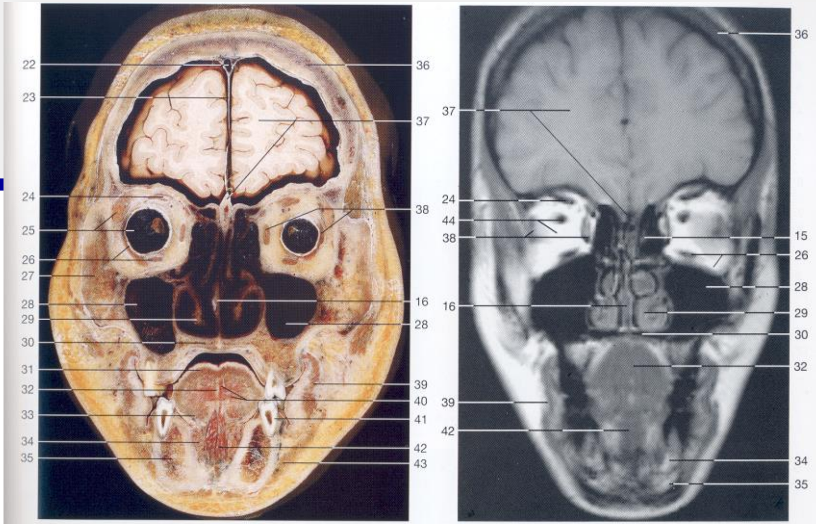
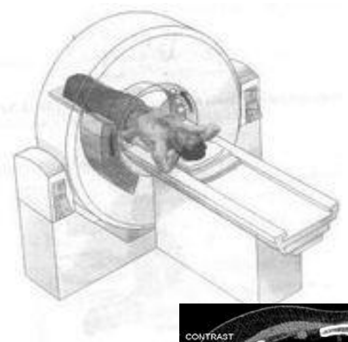
Motivation

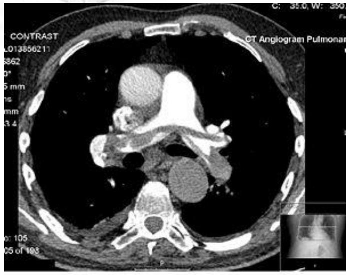
- What does human body look like on the inside?
- Invasive Techniques
  - Surgery, Histology, Biopsy
- Non-Invasive Techniques
  - Medical Imaging Devices
  - Still: One has to consider Radiation (CT, PET)!



Coronal section through the head at the level of the second premolar of the mandible.

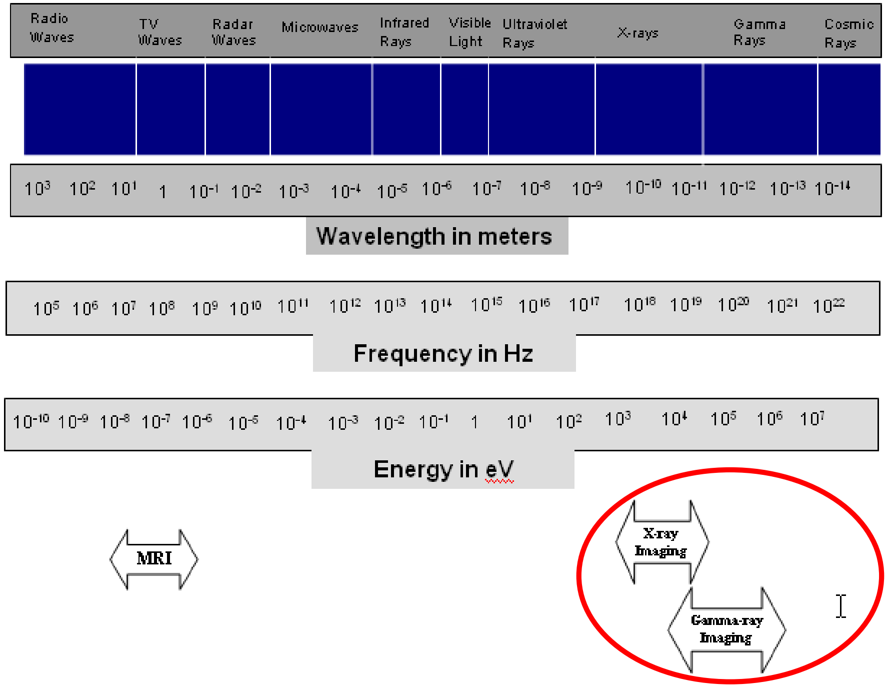
Coronal section through the head (MRI scan, courtesy of Prof. Dr. A. Heuck, Munich, Germany). Note the situation of the head cavities.





X-Ray Computed Tomography

- In 1895, Roentgen noticed „rays of mysterious origin“ from a Crooke‘s tube
- Called them „X-rays“
  - EM-waves of higher frequency than visible light
  - Ionizing radiation



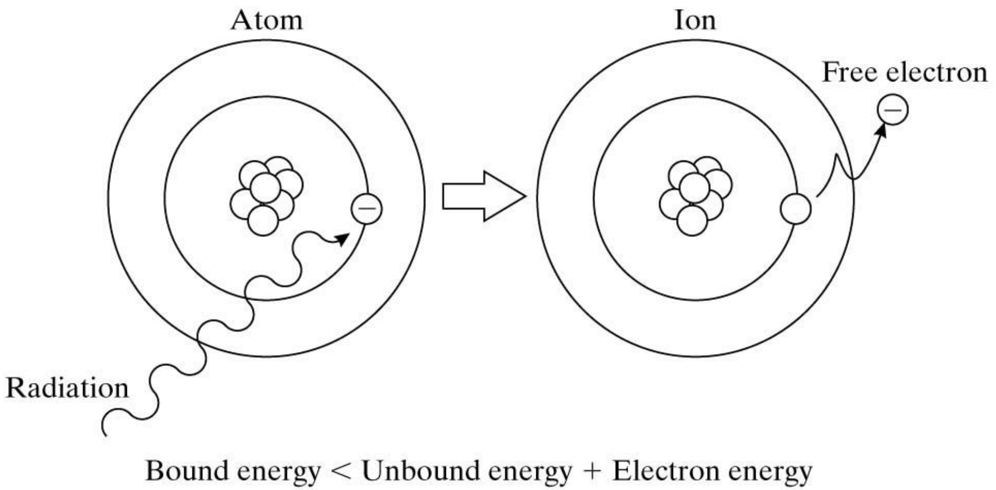
Radiation

- Ionizing radiation: may eject electrons from atom
  - Possesses higher energy than non-ionizing radiation

- Examples:
  - Ionizing
    - Electromagnetic waves, short wavelength (UV light, X-rays, gamma rays)
    - Particulate radiation (protons, electrons, positrons with high kinetic energy)
  - Non-ionizing
    - EM waves, long wavelength (radio, visible light)

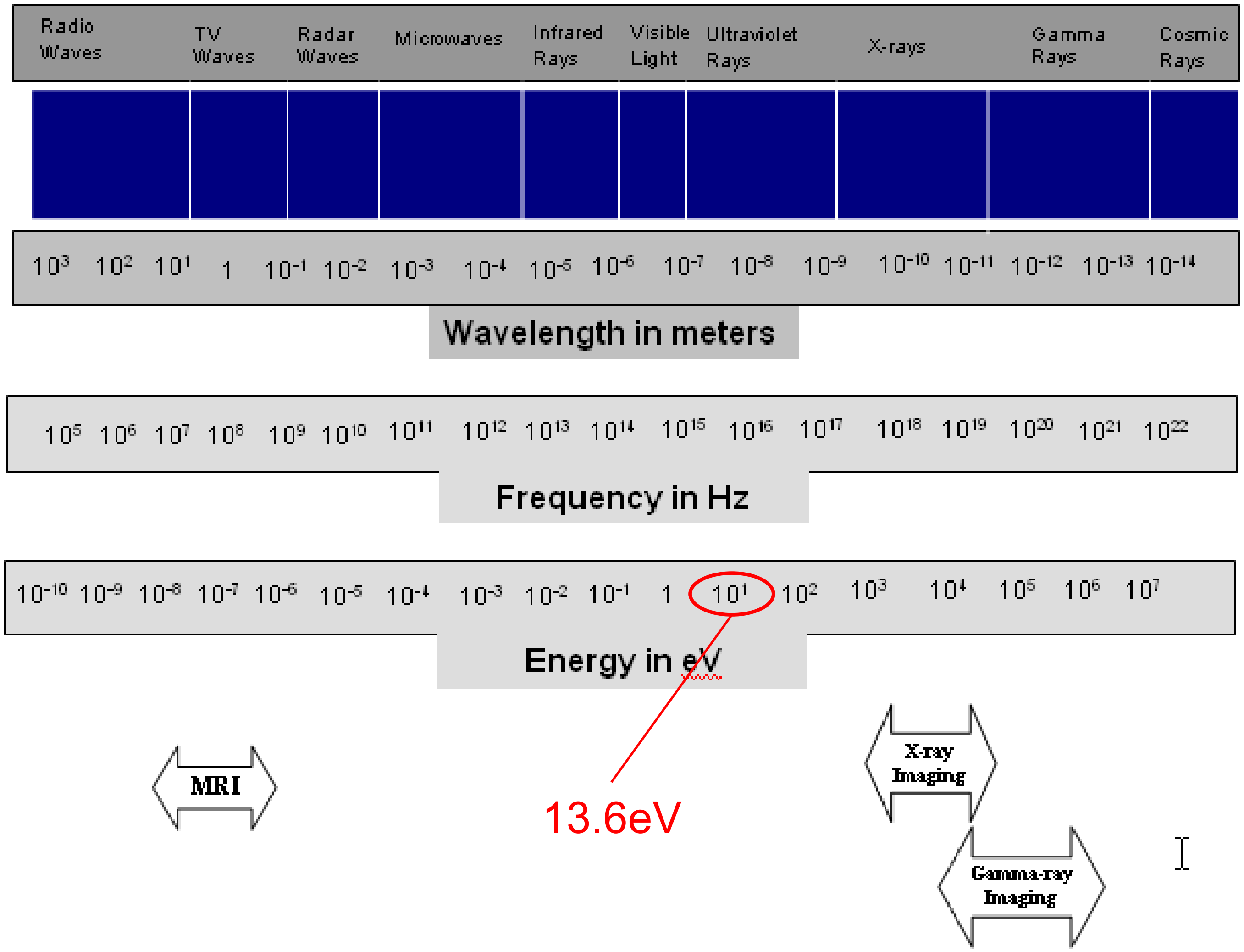
Ionization

- Ionizing radiation transfers energy (>=binding energy) to electron -> ejected from atom (ionization)
- Result: Ion + free electron



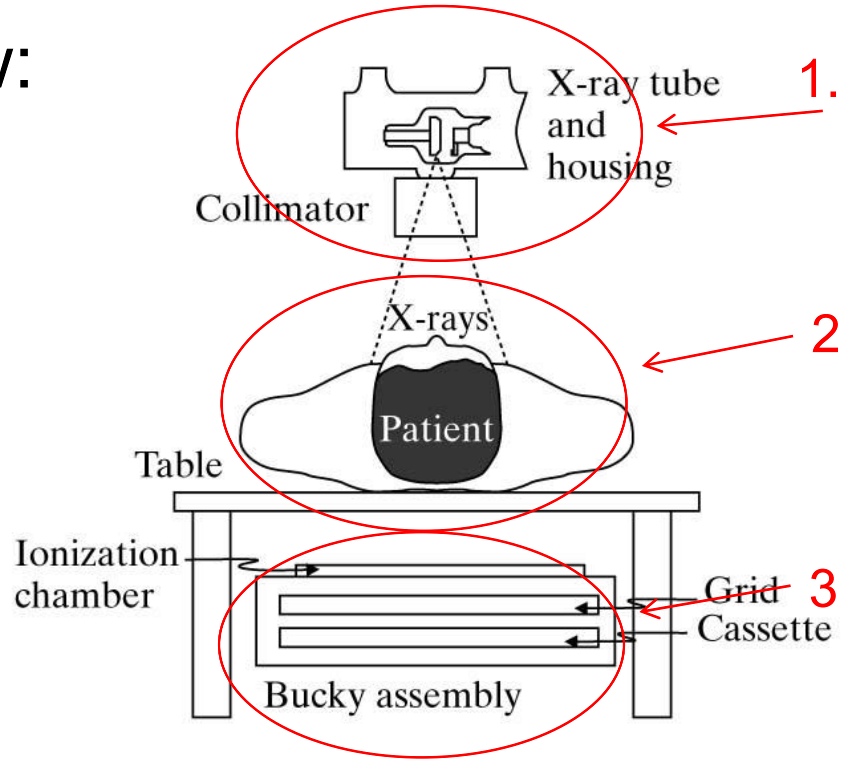
  - Difference is called electron binding energy [eV] (electron volts) e.g. Hydrogen atom: 13.6 eV, Tungsten: mean = 4 keV
  - Binding energy depends on element and shell
  - If electron “holes” are filled again -> Energy is freed (characteristic radiation): e.g. X-ray photon

EM Radiation - Spectrum



X-ray Based Imaging

Overview:



1. Electron Beam creates X-ray Photons (Particle Radiation)
2. X-ray Photon Attenuation depends on Tissue -> provides Signal
3. Detection unit measures incoming X-ray energy

Figure 5.2

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

X-ray Generation

X-Ray Tube

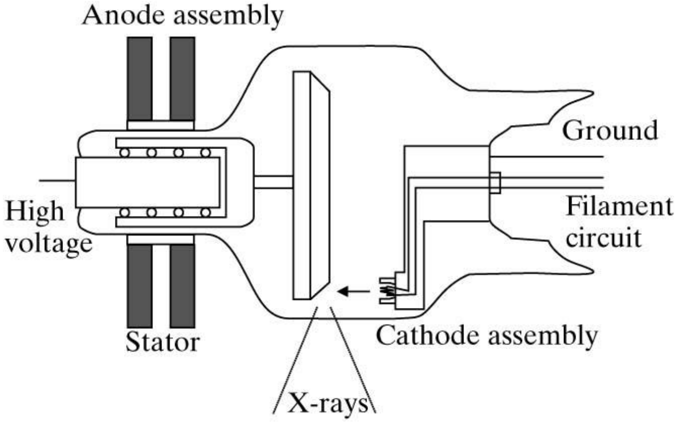


Figure 5.4

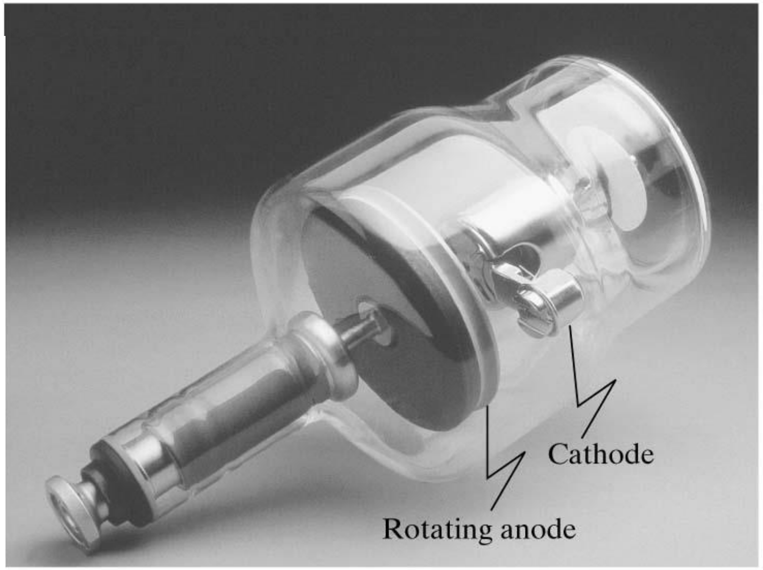


Figure 5.3

X-ray Generation

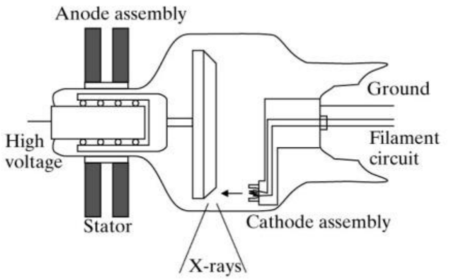


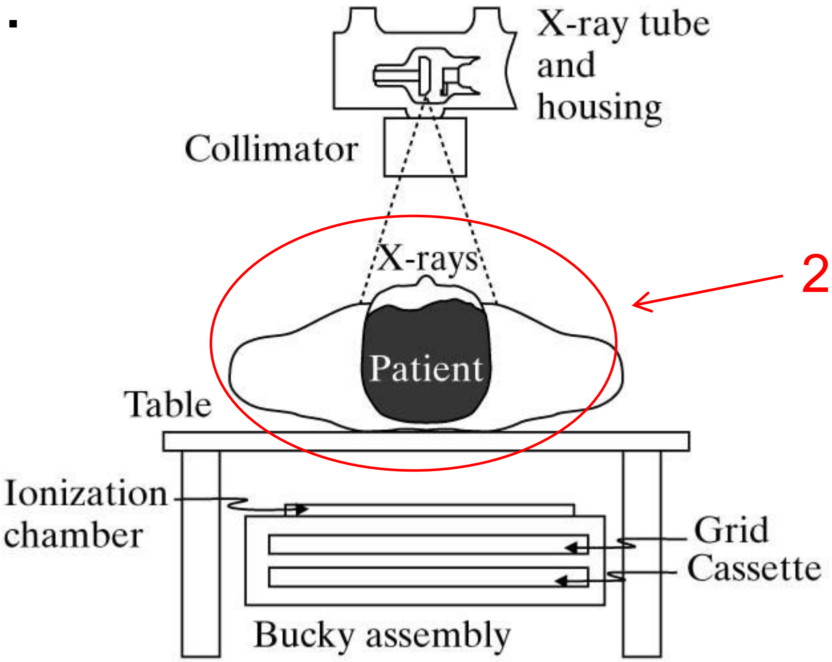
Figure 5.4

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

- Heated filament -> cathode (electrons)
- 30 – 120 kV voltage accelerating electron beam in vacuum
- Electron beam hits target anode
  - Tungsten anode (rotates for cooling)
- X-ray photons emit (mainly Bremsstrahlung)
- Low energy photons filtered (anode, glass body, aluminium foil)
- Collimator photon beam restrictor (direction)

X-ray Based Imaging

Overview:



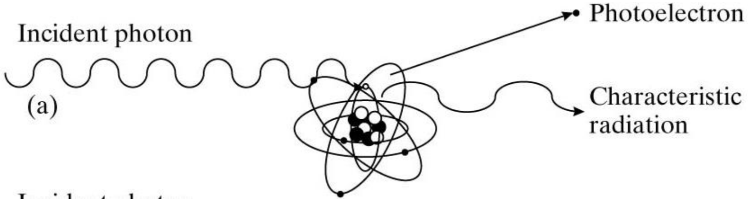
2. X-Ray Photon Attenuation depends on Tissue -> provides Signal

Figure 5.2

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.
ISBN 0-13-065353-5.

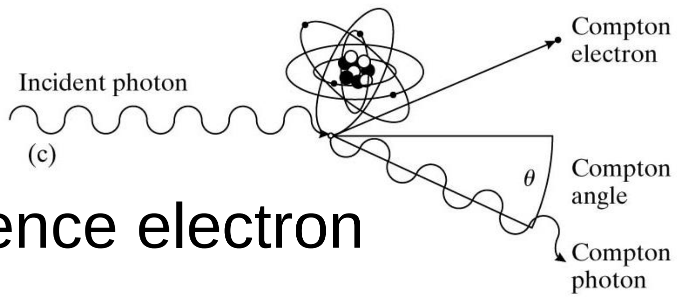
Interaction of X-rays with Matter

- Photoelectric Absorption



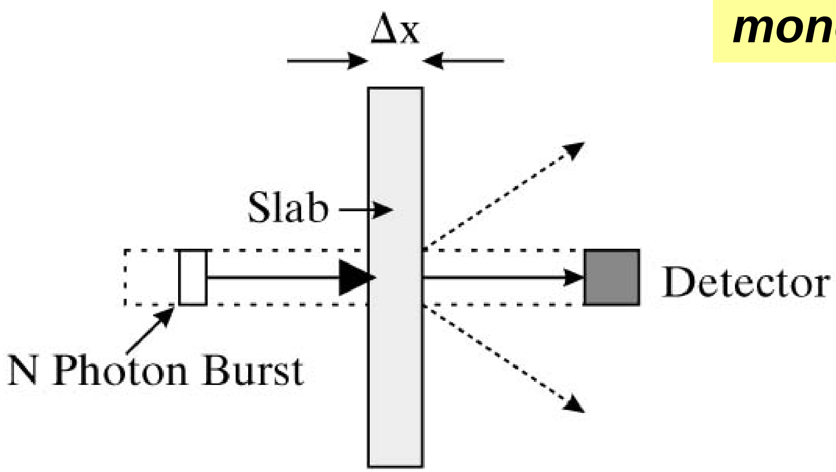
  – Photon loses all of its energy to electrons
  – Attenuation of EM beam (contrast)
  – Generates photoelectrons (particle radiation -> noise)

- Compton Scattering



  – Interaction of photon with valence electron
  – Energy loss & direction change
    (limit in resolution, noise)

Attenuation of EM Radiation



„narrow beam geometry“

monoenergetic photons, homogeneous material:

$$\boxed{\begin{aligned} N &= N_0 e^{-\mu \Delta x} \\ I &= \hbar \nu \frac{N}{A \Delta t} \\ I &= I_0 e^{-\mu \Delta x} \end{aligned}}$$

N_0… # photons at x=0
μ … linear attenuation coefficient
A … area (normal to ray)
t … time
I … intensity of x-ray beam = energy fluence rate
E … energy $E = \hbar \nu$

polyenergetic photons, inhomogeneous material:

$$I(x) = \int I_0(E) e^{-\int_0^x \mu(x', E) dx'} dE$$

Linear Attenuation Coefficient

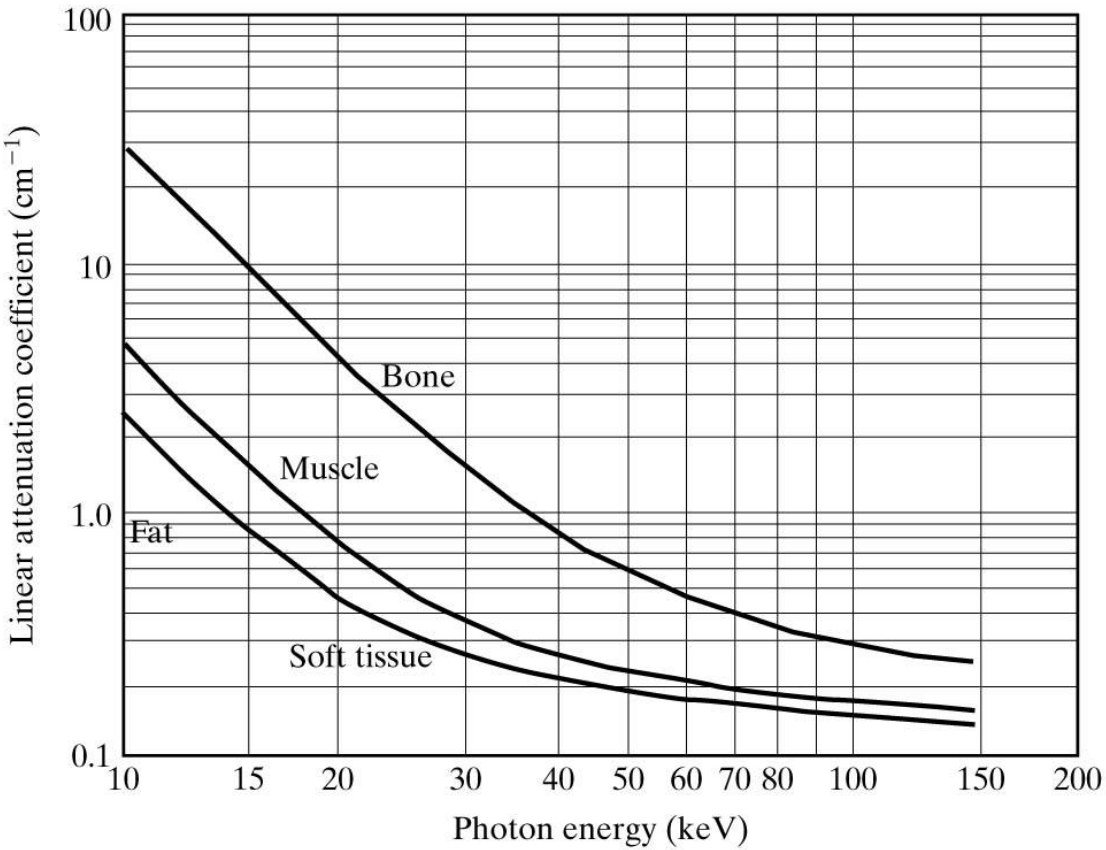


Figure 4.8

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.
ISBN 0-13-065353-5.

… is dependent of material and energy

… is what we want to reconstruct

X-ray Based Imaging

Overview:

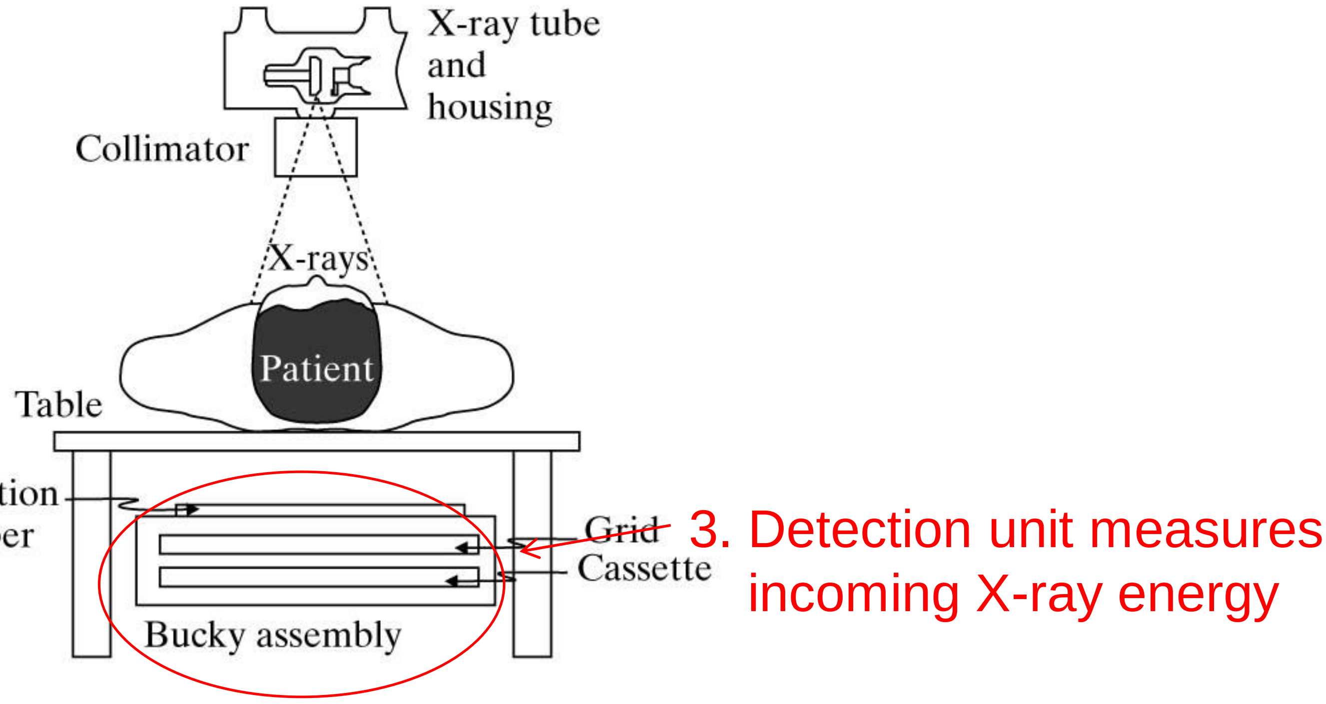


Figure 5.2

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

X-ray Detectors

- Digital detectors
  - Scintillation crystals (visible light from radiation)
  - Photomultiplier

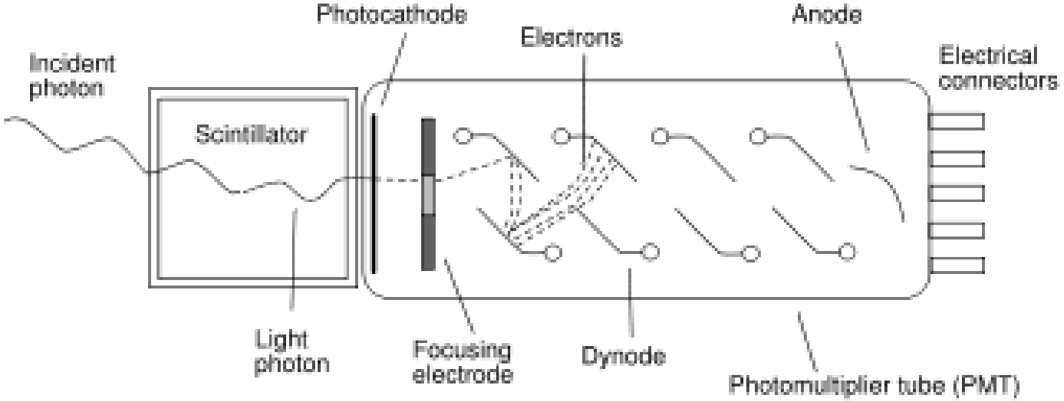
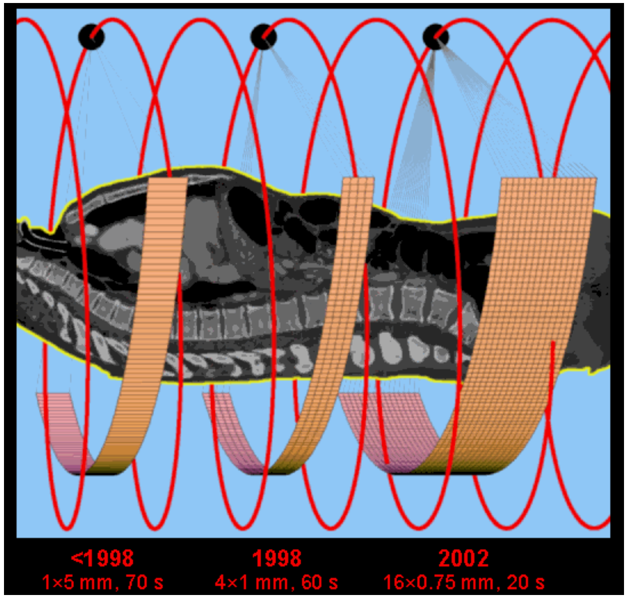


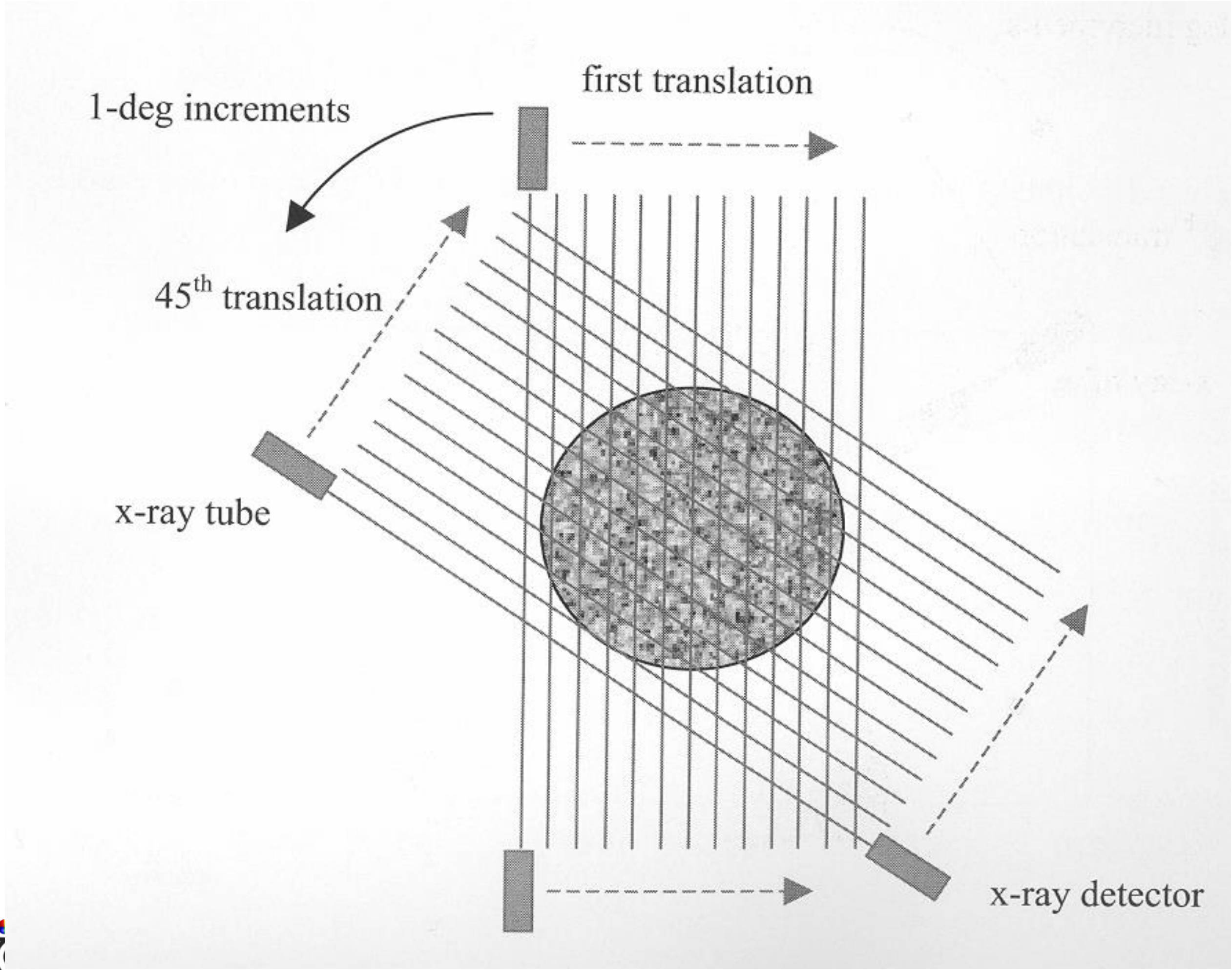
Image taken from Wikipedia

Computed Tomography (CT)

- Principle: X-ray transmission
- 3D object → 3D image (volume)
- Measurements (projections) are *computed* into images of cross-sections
  - Reconstruction from projections
- Types:
  - (Standard) single-slice CT
  - Multi-slice CT (several detector rows, very fast!)



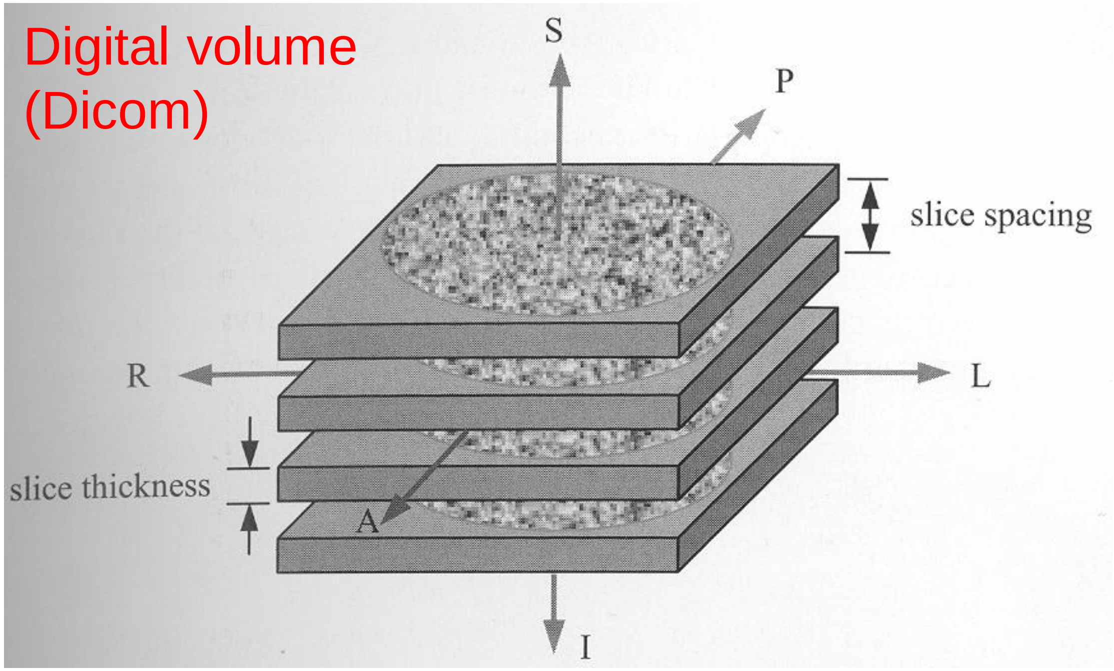
1st Generation (G) CT Scanner Geometry



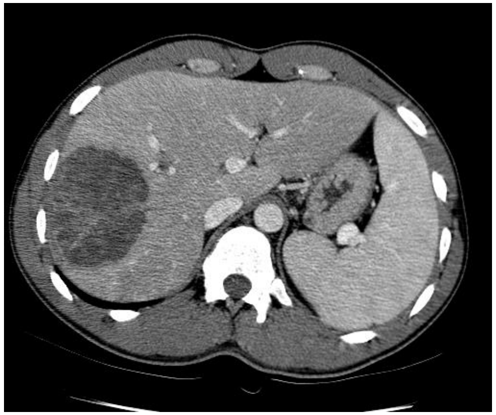
- single source & detector
- parallel motion & rotation
- arbitrary # rays
- no scattering!
- slow

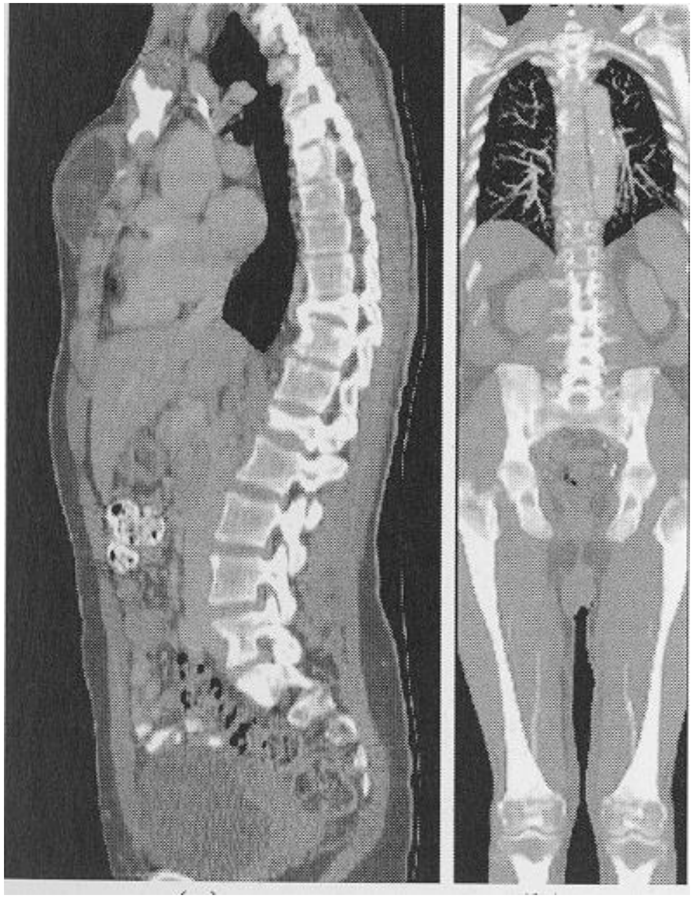
- today: 7G helical multislice scanner

CT Slice Stack (Volume)



axial

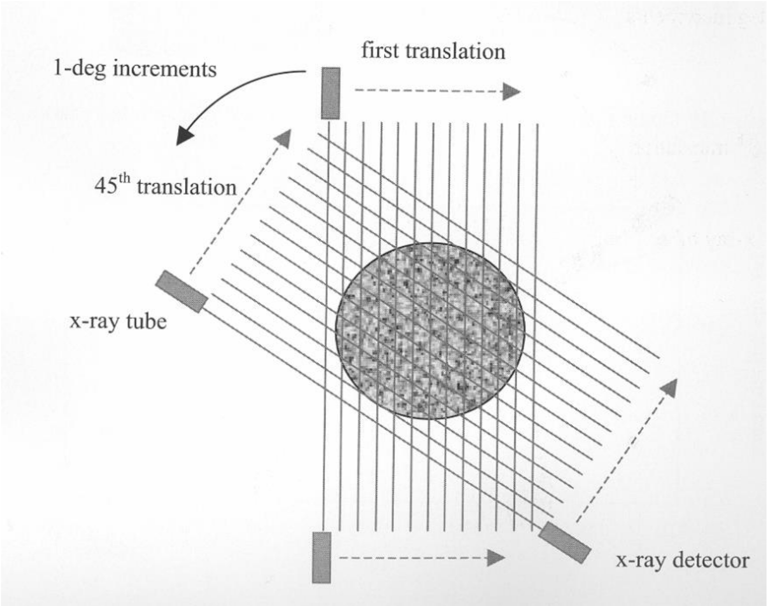




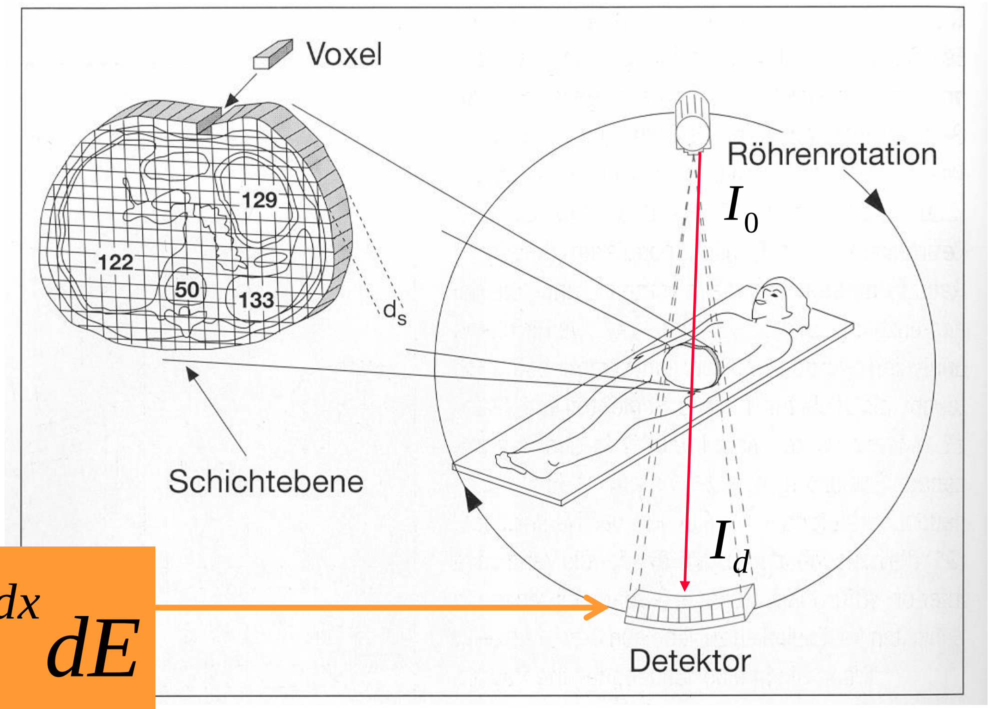
sagittal coronal

Image Formation – Line Integrals

- Q: How can we reconstruct a cross section (slice) ?



- 1D Projections of a Slice – Radon Transform
- Compute Inverse Radon Transform



polyenergetic source:

$$I_d = \int I_0(E) e^{-\int \mu(x,E)dx} dE$$

Image Formation – Line Integrals

- Integral over energy is intractable for image reconstruction!

$$I_d = \int I_0(E) e^{-\int \mu(x',E)dx'} dE$$

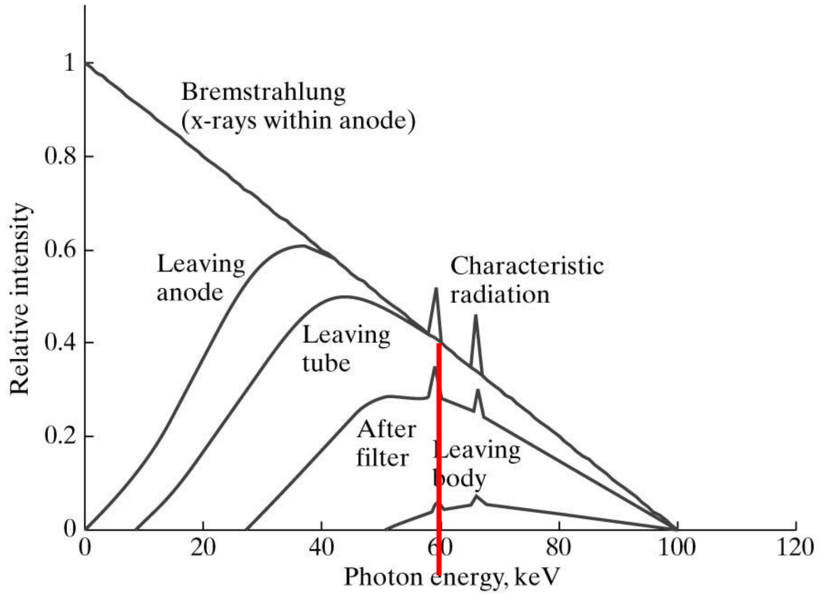


Figure 5.5

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

Image Formation – Line Integrals

- Integral over energy is intractable for image reconstruction!

$$I_d = \int I_0(E) e^{-\int \mu(x',E)dx'} dE$$

- Assumption: effective energy $\bar{E}$ is used, defined as the energy that, in a given material will produce the same measured intensity from a monoenergetic source as is measured using the actual polyenergetic source.

$$I_d = I_0(\bar{E}) e^{-\int_0^d \mu(s,\bar{E})ds}$$

Image Formation – Line Integrals

I_0 and I_d can be measured easily!

Basic projection measurement:

$$g_d = -\ln\left(\frac{I_d}{I_0}\right) = \int_0^d \mu(s,\bar{E})\,ds$$

… is a line integral of the linear attenuation coefficient at the effective energy of the scanner

The reference intensity I_0 must be measured for each detector → calibration step

CT Numbers

- CT reconstruction: μ value for each pixel (voxel) of slice
- Problem: different CT scanners → different tubes → different effective energy → same object → different numerical values of μ
- Also: Replacement of x-ray tubes necessary!
- Solution: CT numbers:

→ comparable results

$$h = 1000 \times \frac{\mu - \mu_{water}}{\mu_{water}} \qquad [HU]$$

HU … Hounsfield Units

h=0 HU for water
h=-1000 HU for air ($\mu = 0$)
h~1000 HU for bone
h~3000 HU for metal, contrast agents

CT Numbers & Gray-value Range

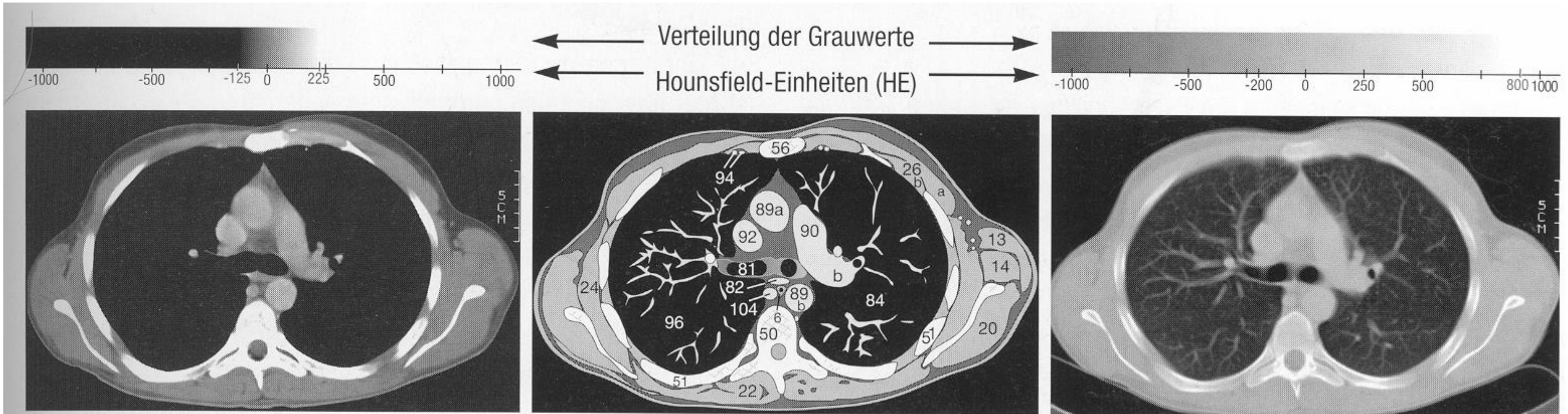


Abb. 13.1a: Weichteilfenster

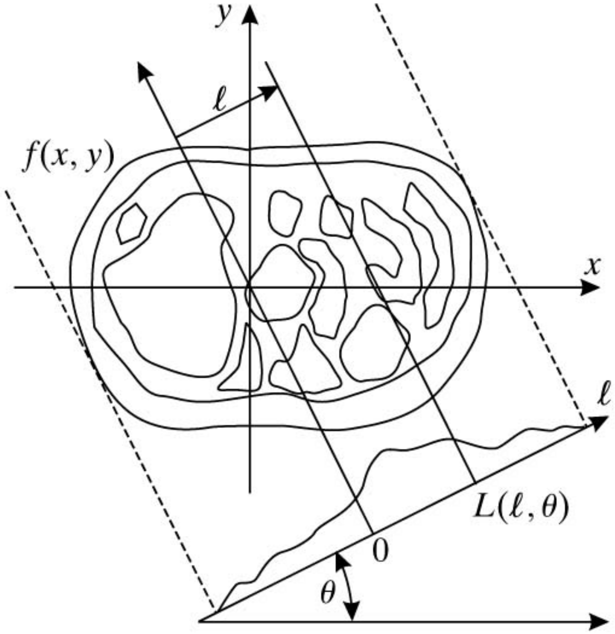
Abb. 13.1b

Abb. 13.1c: Lungenfenster

Parallel-Ray Reconstruction

- Goal: CT numbers h over the entire cross-section

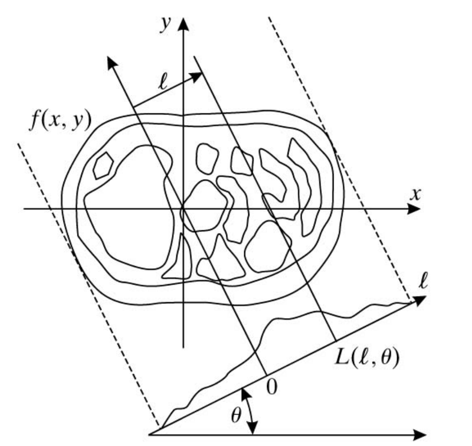
  (→ image)
- Q: How can we reconstruct μ given a collection of its line integrals?

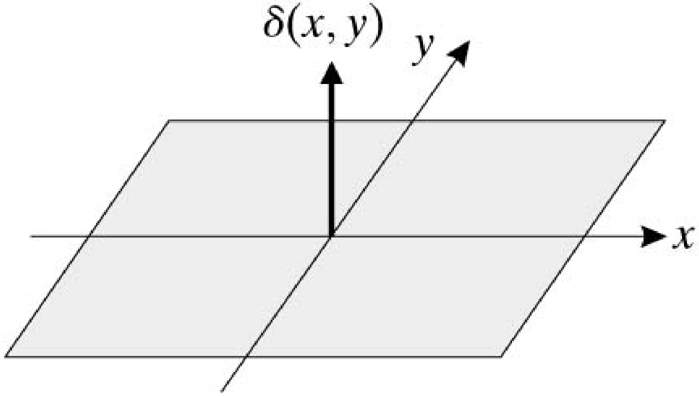


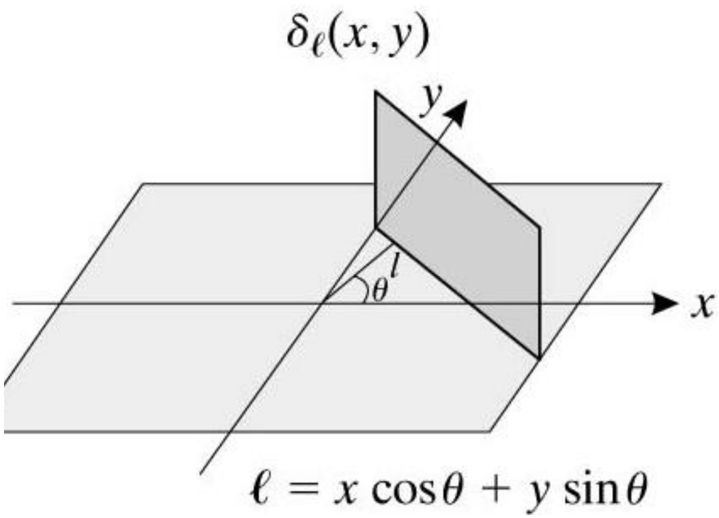
EQ line:

$$L(l,\theta)=\{(x,y)|x\cos\theta+y\sin\theta=l\}$$

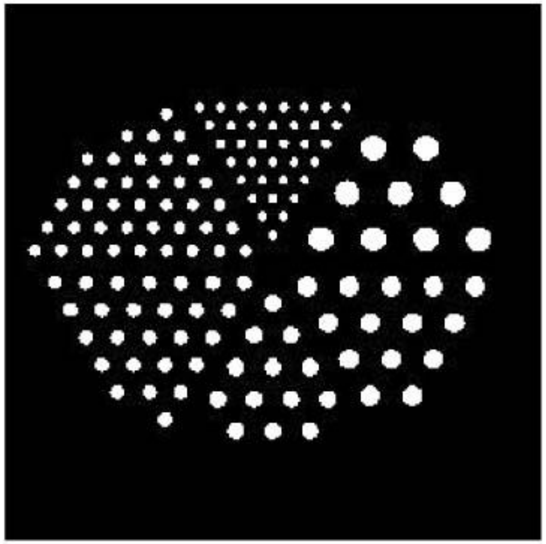
Line Integral

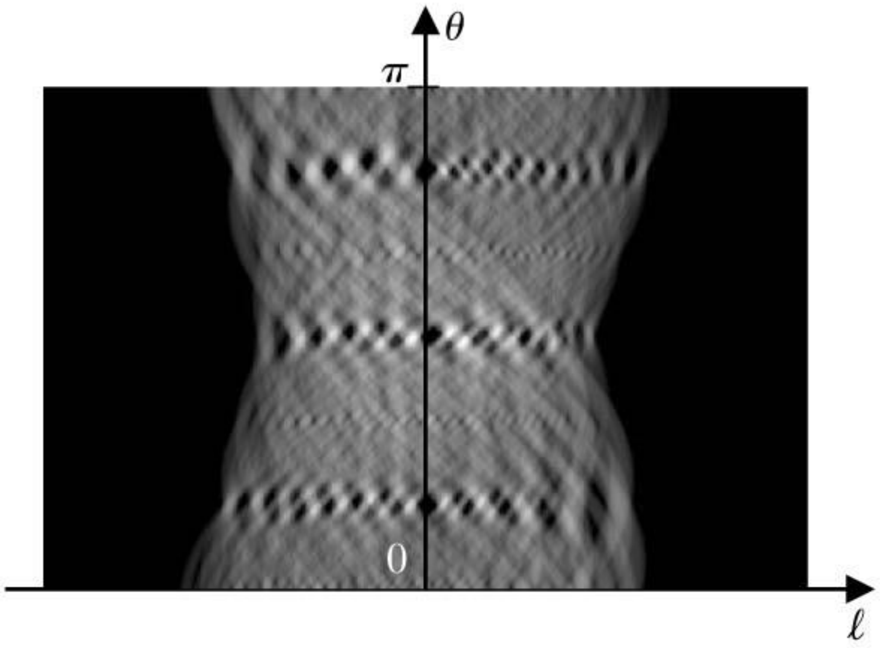






2D Radon Transform - Sinogram





Object

Sinogram of the Object

An image of $g(l, \theta)$ with l and θ as rectilinear coordinates is called a sinogram.

Matlab

Reconstruction - Backprojection

- Single projection → infinite number of $f(x, y)$

  → more projections are needed!

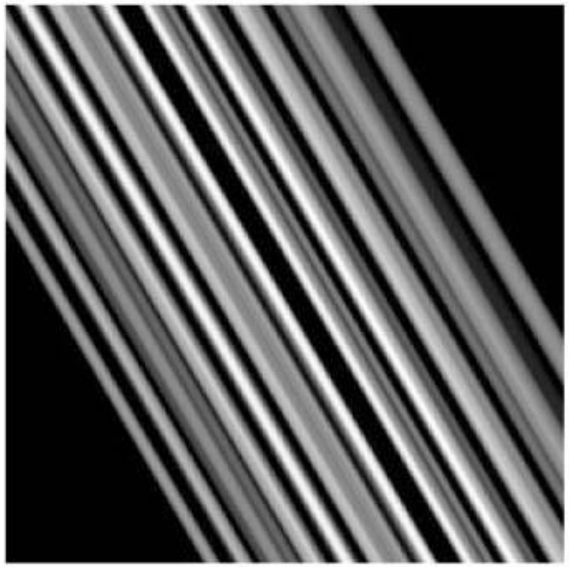
Simple: Backprojection image:

$$b_\theta(x, y) = g(x\cos\theta + y\sin\theta, \theta)$$

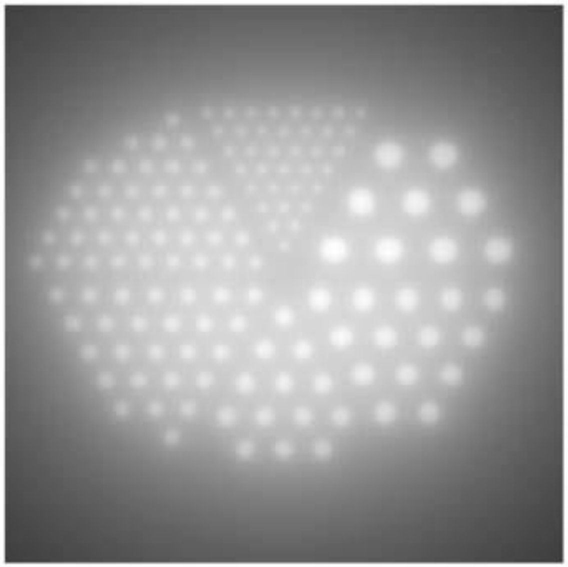
Backprojection summation image (laminogram):

$$f_b(x, y) = \int_0^\pi b_\theta(x, y)\,d\theta$$

Backprojection Summation Image



One backprojection image $b_{30°}$

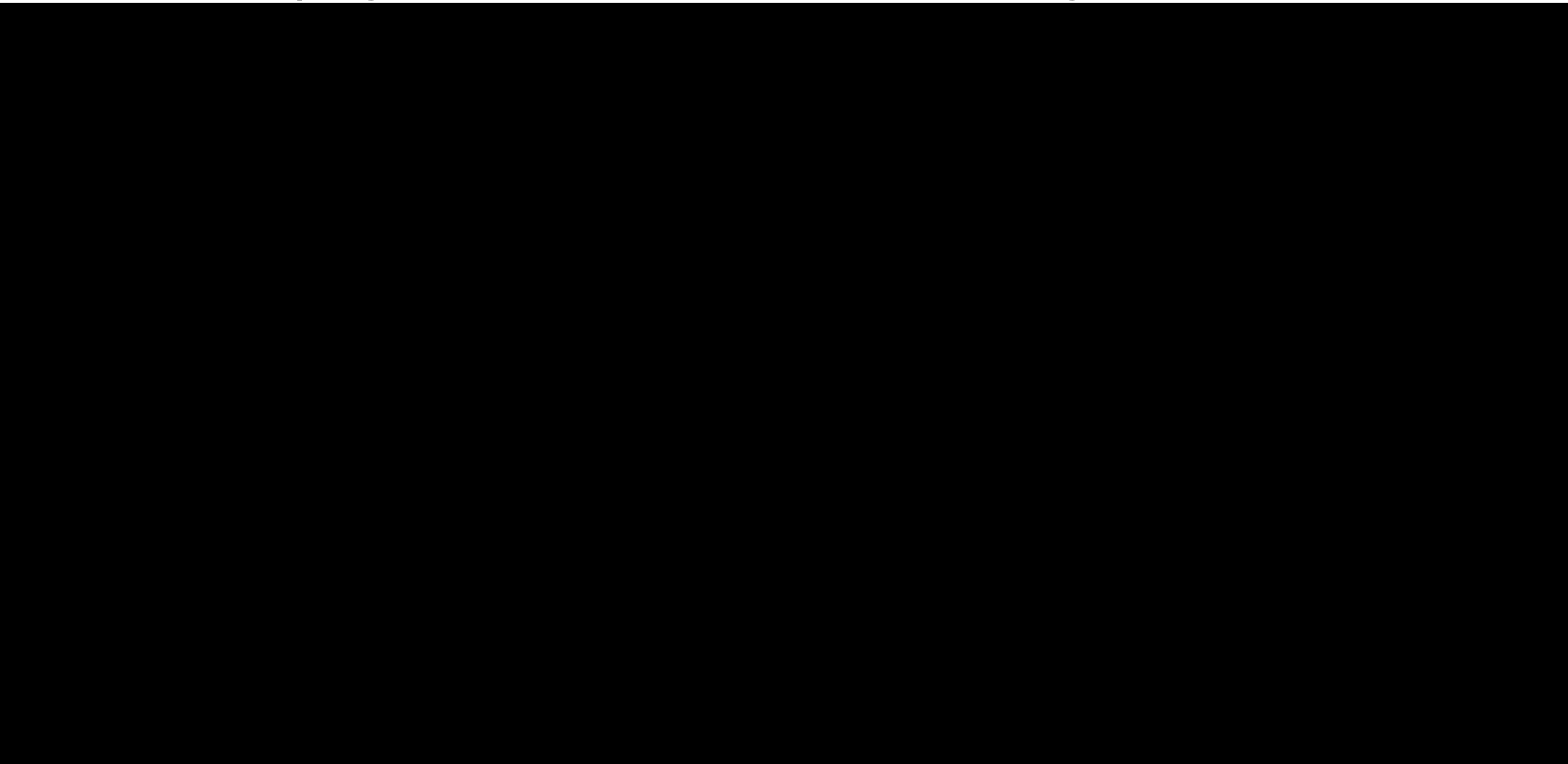


Backprojection summation image

Better approaches exist (see blurriness in summation image)

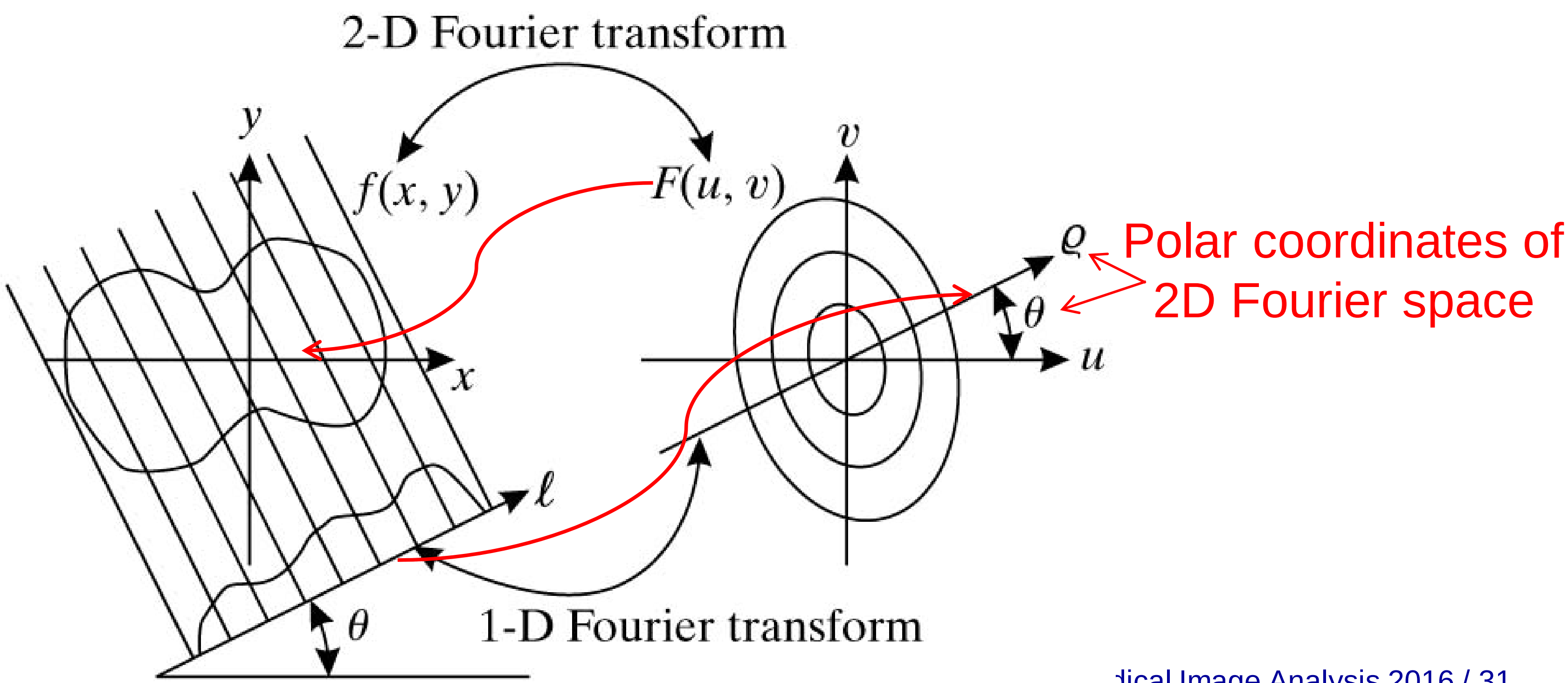
Projection-Slice Theorem

- Very important relation between 1D Fourier transform of a projection and 2D transform of object



Projection-Slice Theorem

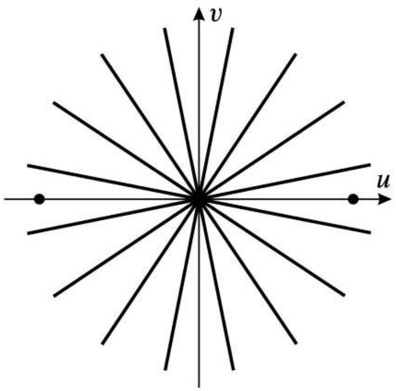
1D FT of a projection equals a line passing through the origin of the 2D FT of the object @ the projection angle!



The Fourier Method (Reconstruction)

- Take the 1D FT of each projection, insert it with the corresponding correct angular orientation into the correct place of the 2D Fourier plane, and take the inverse 2D FT of the result.

$$f(x,y) = FT_{2D}^{-1}\{G(\rho,\theta)\}$$



Method is not widely used in CT (sampling, interpolation)!

Filtered Backprojection

Inverse FT
in polar coord.:

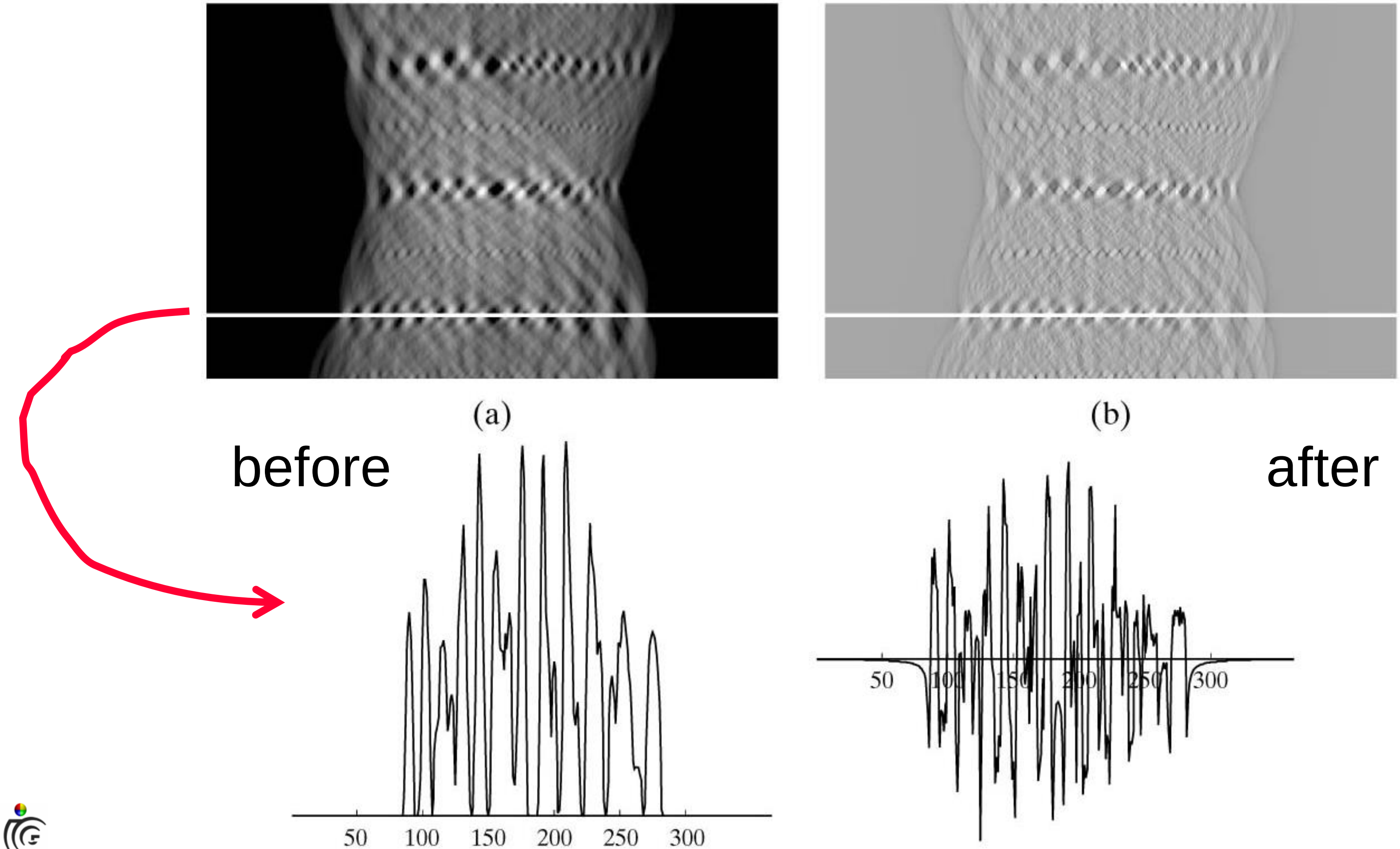
$$f(x,y)=\int_0^{2\pi}\int_0^{\infty}F\left(\rho\cos\theta,\rho\sin\theta\right)e^{j2\pi\rho\left(x\cos\theta+y\sin\theta\right)}\rho d\rho d\theta$$

Reconstruction in Three Steps

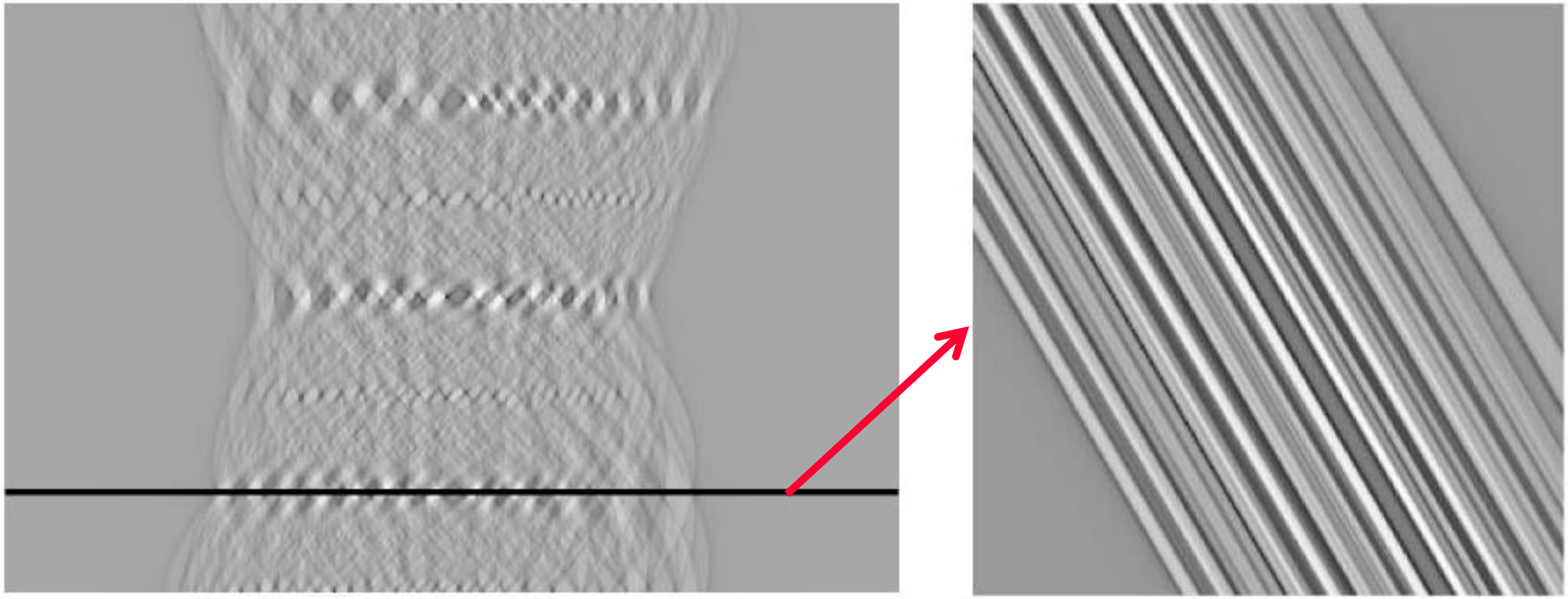
Input *Sinogram*:

1. Filtering
2. Backprojection
3. Summation

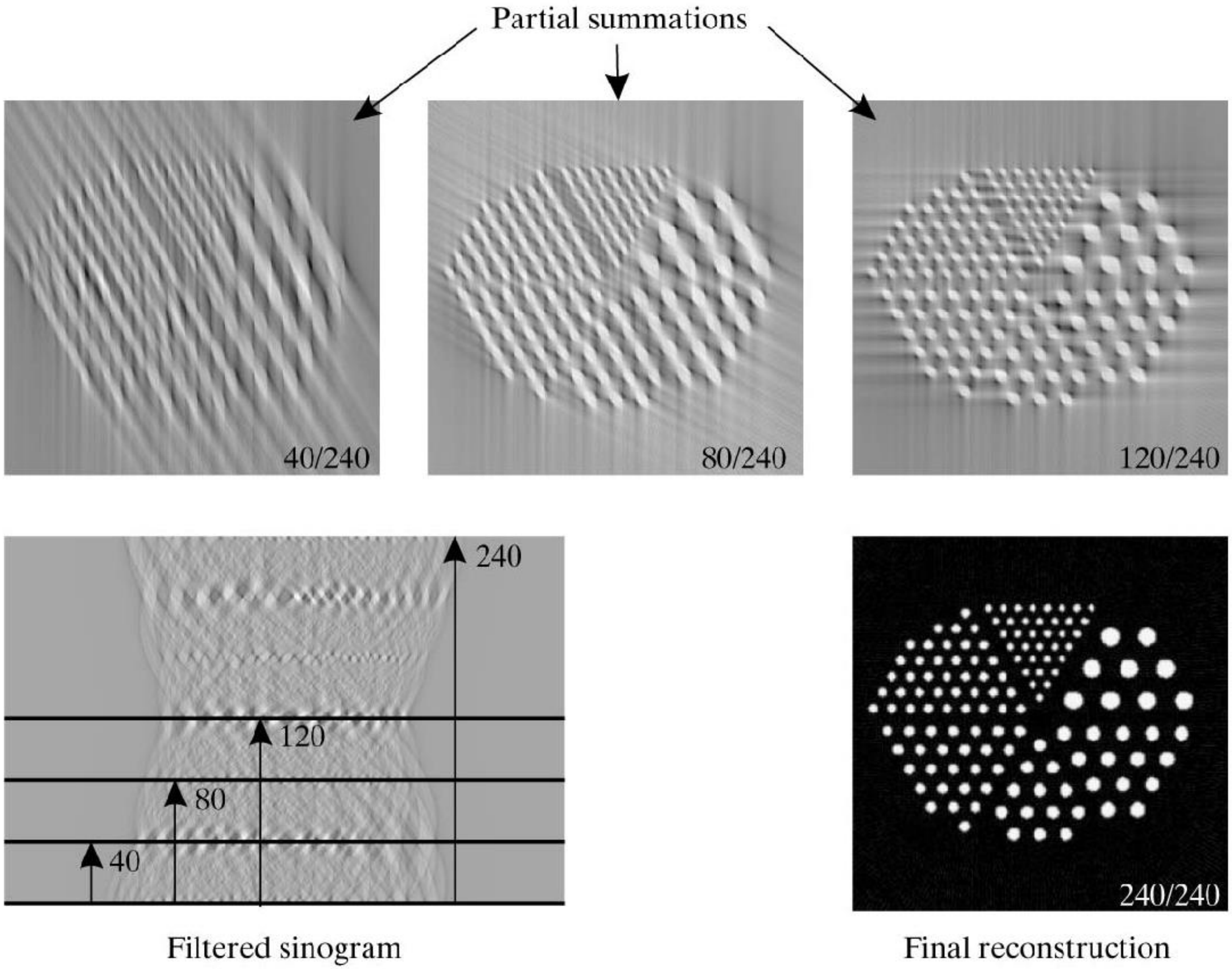
Convolution Step (Filtering)



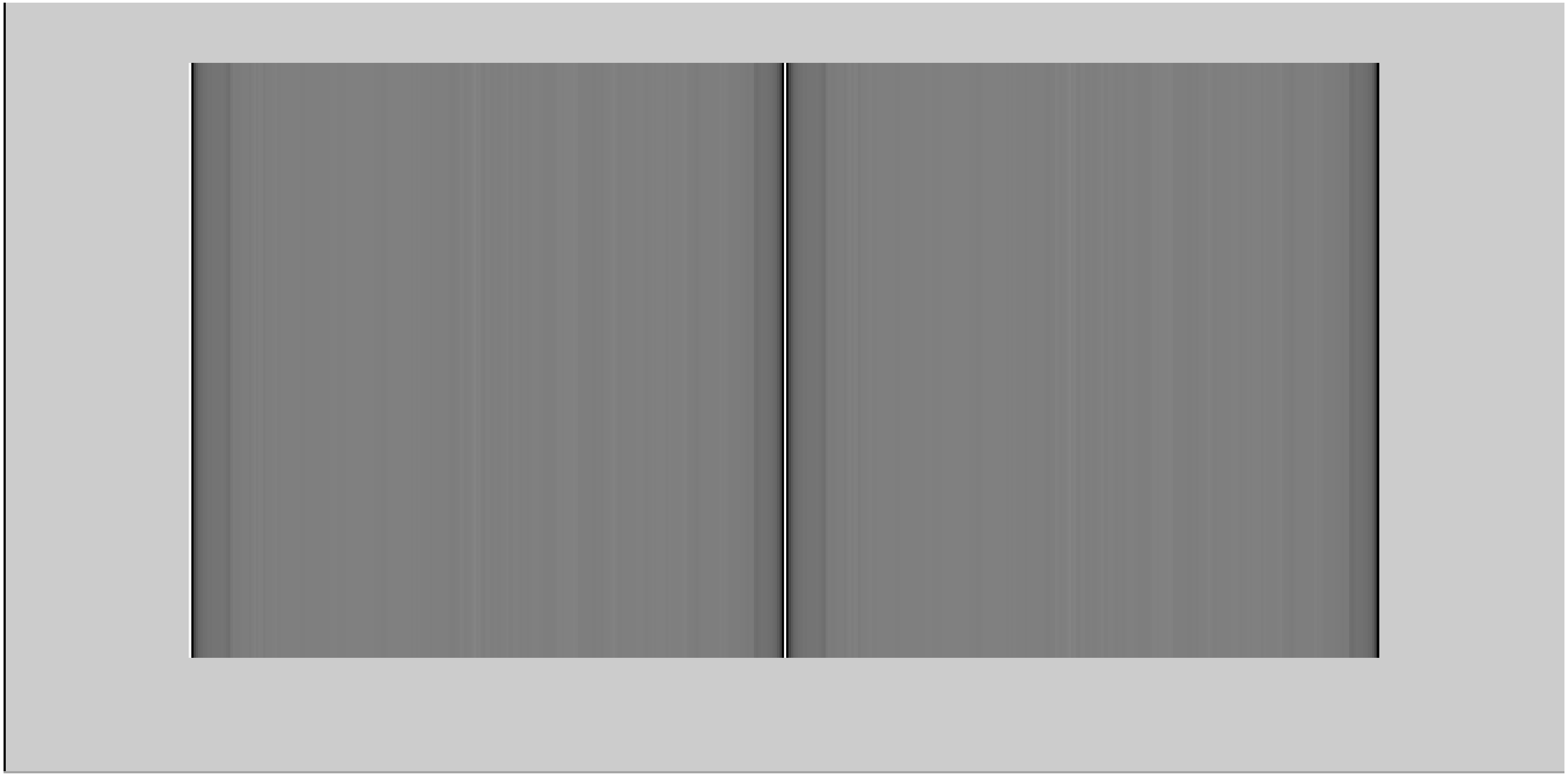
Backprojection Step



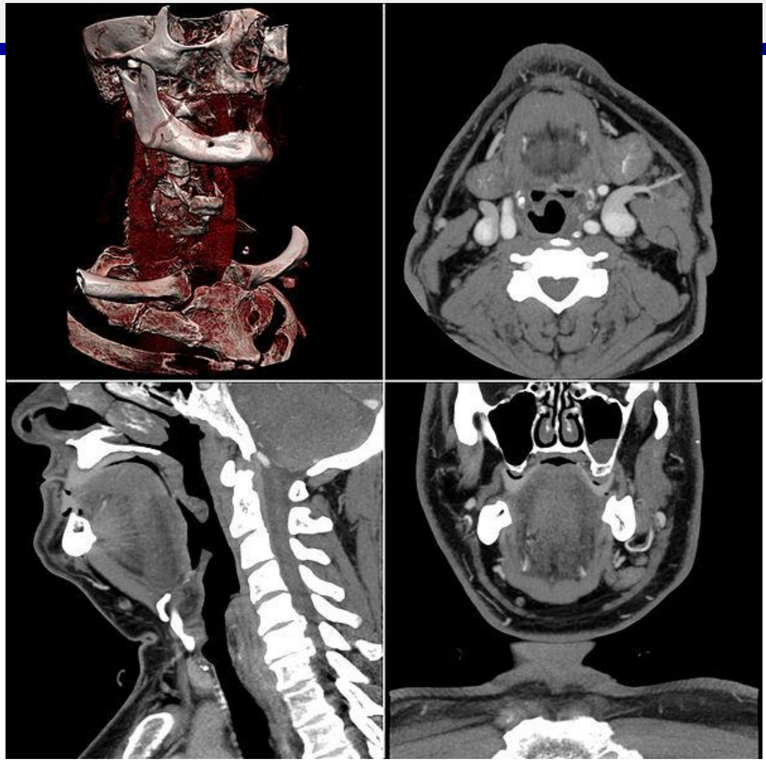
Summation Step



Filtered Backprojection - Movie



Conclusion – Development Trends

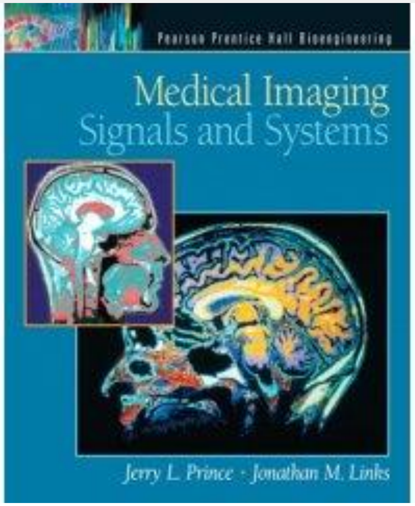


- Faster CT scans
- Higher resolution / more detail
- Iterative, algebraic reconstruction
- Better image quality

Amount of data is "exploding" → need for **automated medical image analysis**

END

Material:



One minute paper:

a) What did I learn today?

b) Which topics remained open?

See you next week!

Homework video

https://www.youtube.com/watch?v=r2UqlnVPX30

Denoising, deconvolution and computed tomography using total variation penalty

Answer three questions till next lecture:

1. *Based on an input signal x, how may one write compactly the output signal y after passing a communication channel h?*
2. *What is the meaning of matrix A in the shown model for CT reconstruction?*
3. *What is the formula for total variation of a 3-dimensional volume? (Integral on R3, using partial derivatives!)*