2020 NCC 의학물리 아카데미 교육프로그램

Physics of CT Basic Principle

Haijo Jung, Ph.D
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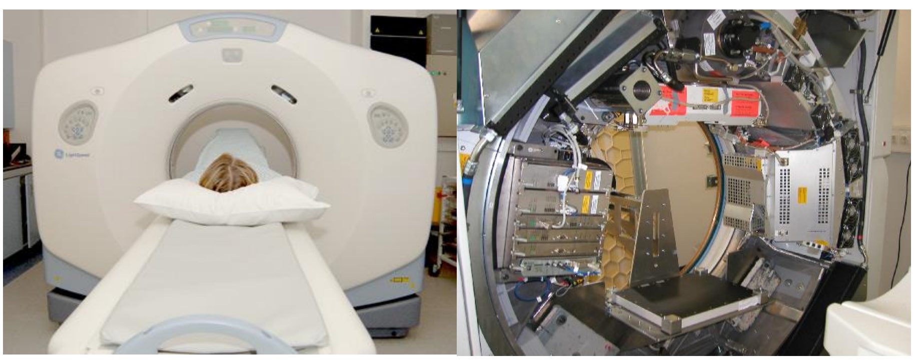
haijo5864@naver.com

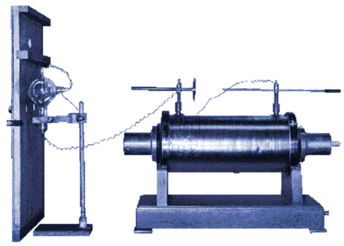
CT (Computed Tomography)



Discovery of X-ray

- Wilhelm Conrad Röntgen (Germany) discovered the generation of new radiation during cathode ray tube experiment by using Crookes Tube in Nov. 08, 1895.
- This radiation have the characteristics of fluorescence, sensitization of film, penetration of opaque objects.
- Röntgen called this radiation with x-ray.
- Novel prize in physics field in 1901.





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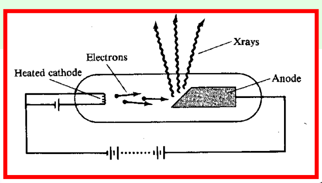
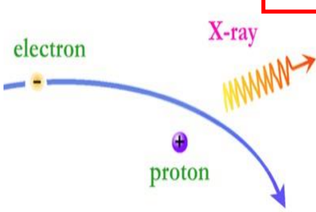
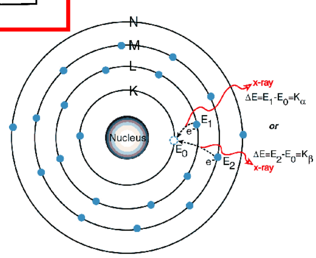
Physics of X-ray Generation

- **Principle of X-ray generation**
  - *Bremsstrahlung X-rays*,
  - *Characteristic X-rays*.







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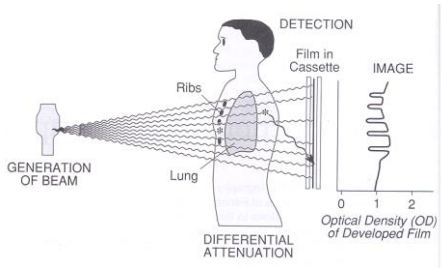
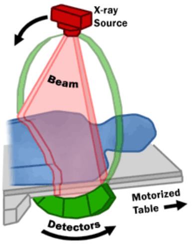
General X-ray Image (Radiography)

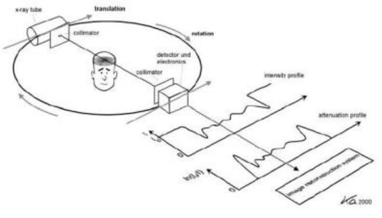
- X-rays are exposed through three dimensional objects. → Penetrated x-rays are recorded on film or detector as two dimensional image
- Lots of information are disappeared due to overlap, the tissues having minute absorption coefficient is hard to discriminate, and scattering x-rays cause bad influences on imaging formation.



CT (Computed Tomography)???



Tomo = "a cut " or "section"
Cross-Sectional Image



History of CT

History

1895 Röntgen

Discovery of X-rays

1917 Radon

Mathematical foundations

1963 Cormack

X-ray tomography idea

1972 Hounsfield

First CT scanning system

1989 Kalender

Introduction of helical CT

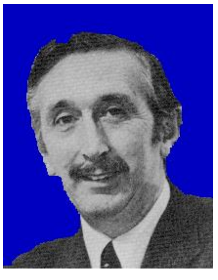
1998

First multi-line CT system

2004

Cone beam CT systems

CT Invention



- X-ray CT scanner was invented by Godfrey N. Hounsfield and James Ambrose (1970)
- The first clinical CT image (1972)
- Novel Price (1979)

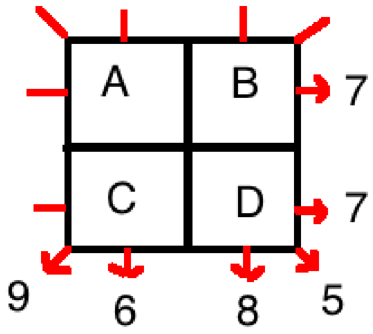
Siemens SIRETOM (1974)

- Acquisition time: 7 min.,
- Image matrix: 80x80 pixels,
- Scan field: 25 cm ,
- Spatial resolution: 1,3 mm (4 lp/cm)

Radon's Theorem

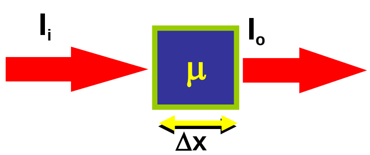
In geometry, **Radon's theorem** on convex sets, published by Johann Radon in 1921, states that any set of *d* + 2 points in R^d can be partitioned into two disjoint sets whose convex hulls intersect. A point in the intersection of these convex hulls is called a **Radon point** of the set.

Johann Radon (1887- 1956) was an Austrian mathematician who is known for Radon-Nikodym theorem and Radon measures.



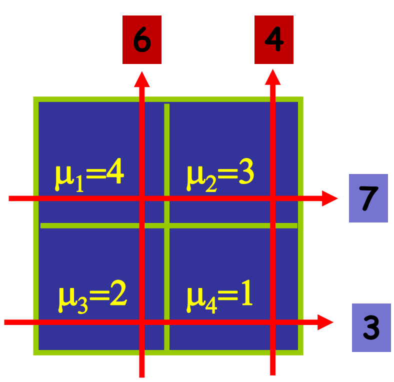
Ans: 2543

Reconstruction Idea



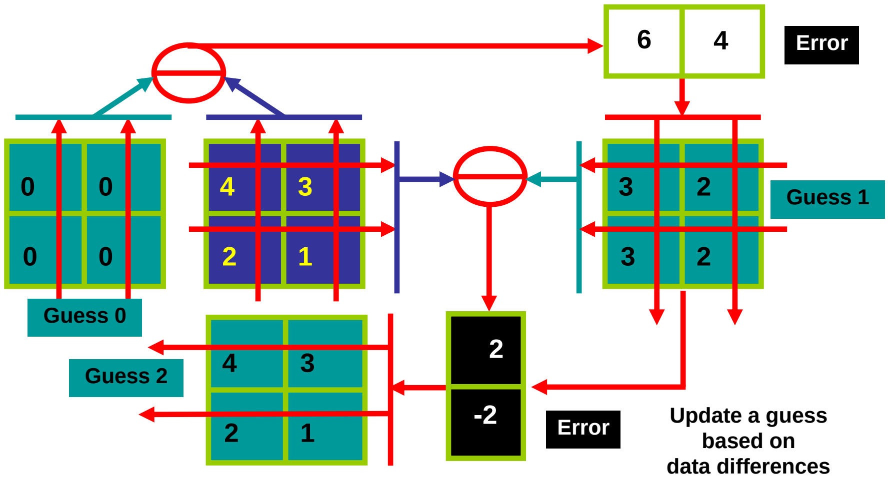
$$I_o = I_i e^{-\mu \Delta x}$$

I_i: input intensity of X-ray
I_o: output intensity of X-ray
μ: linear X-ray attenuation



$$\begin{cases} \mu_1 + \mu_2 = 7 \\ \mu_3 + \mu_4 = 3 \\ \mu_1 + \mu_3 = 6 \\ \mu_2 + \mu_4 = 4 \end{cases}$$

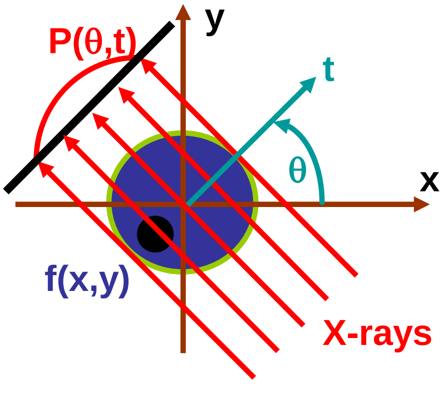
Algebraic Reconstruction Technique

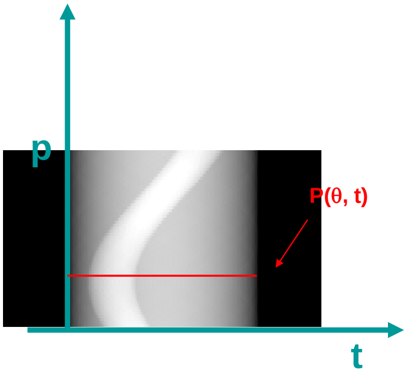


Projection & Sinogram

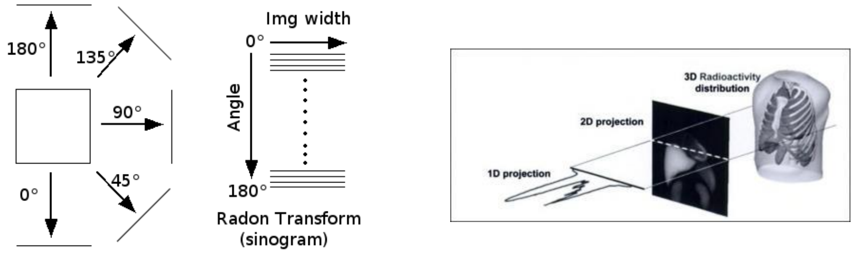
Projection: All ray-sums in a direction

Sinogram: All projections





What is Sinogram?



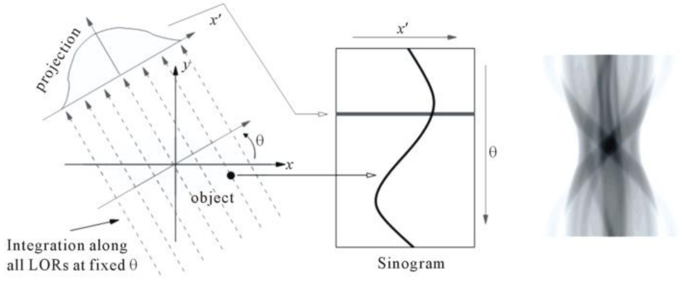


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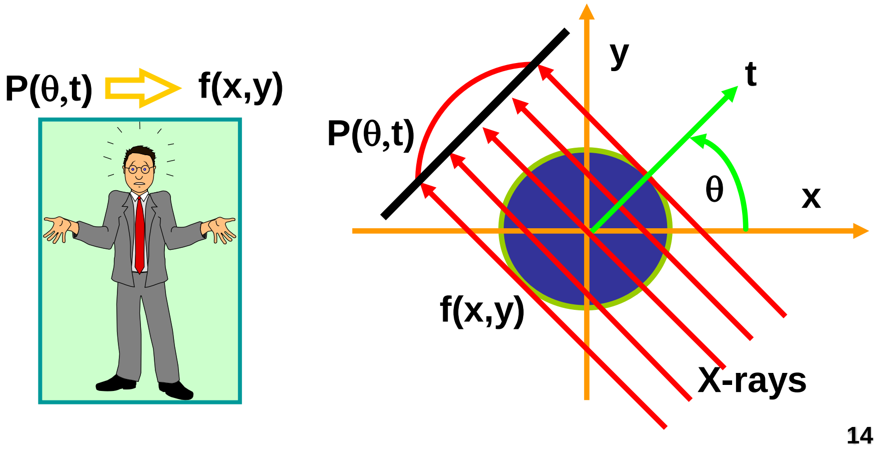
Image Reconstruction

Image reconstruction from projections



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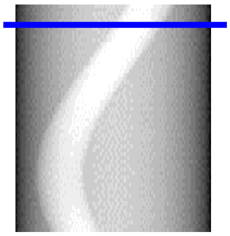
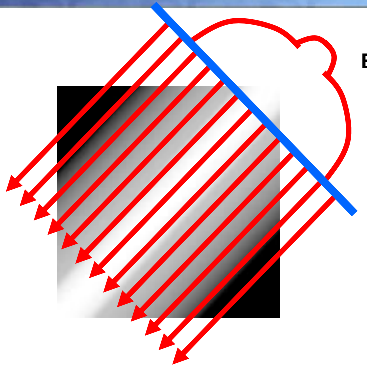
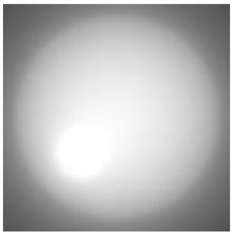
Backprojection

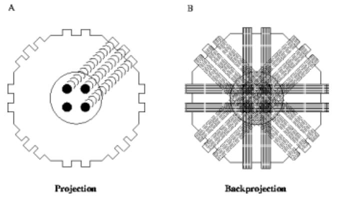
Sinogram

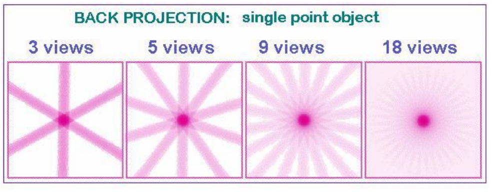




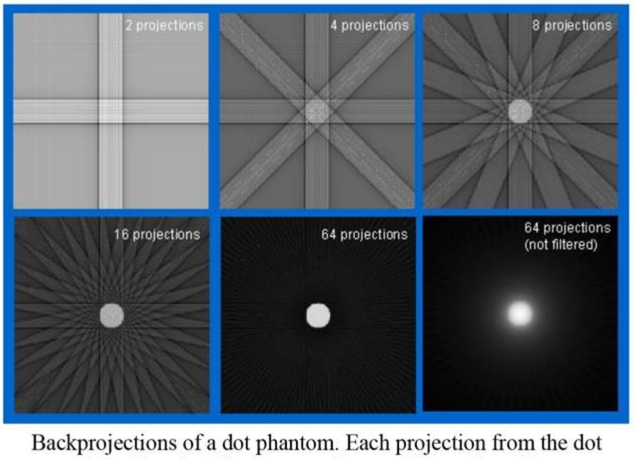
Backprojected Image





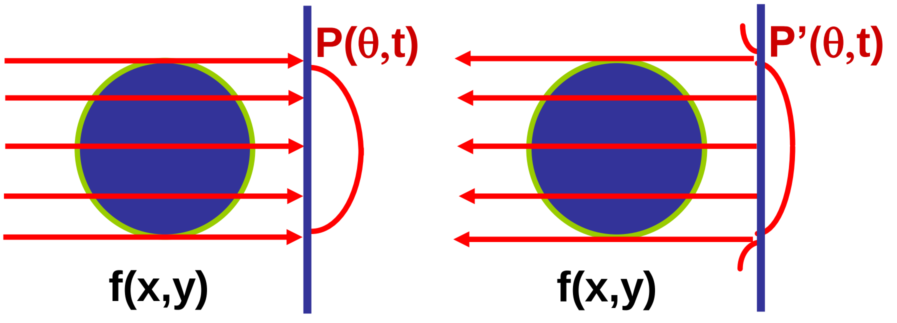


Projection & Backprojection



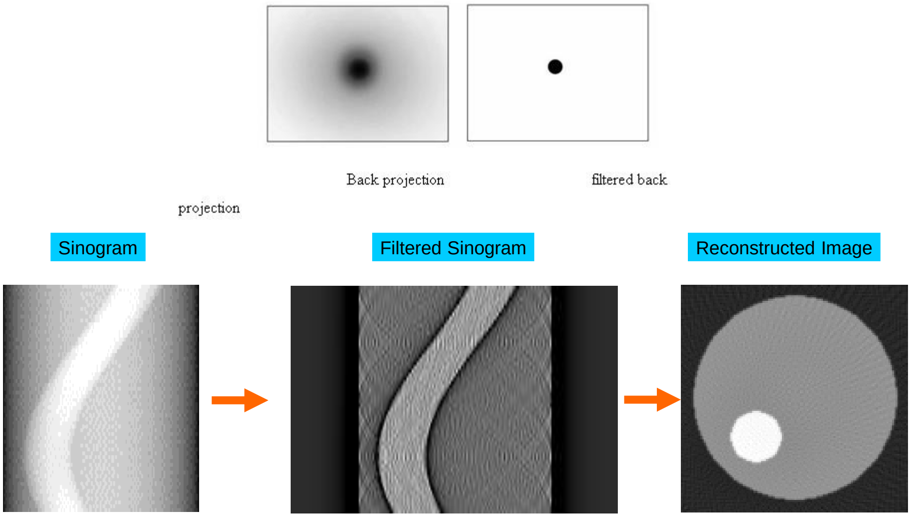
Backprojections of a dot phantom. Each projection from the dot is backprojected, or smeared across the section. The backprojections are added together, resulting in a reconstruction.

Filtered Backprojection

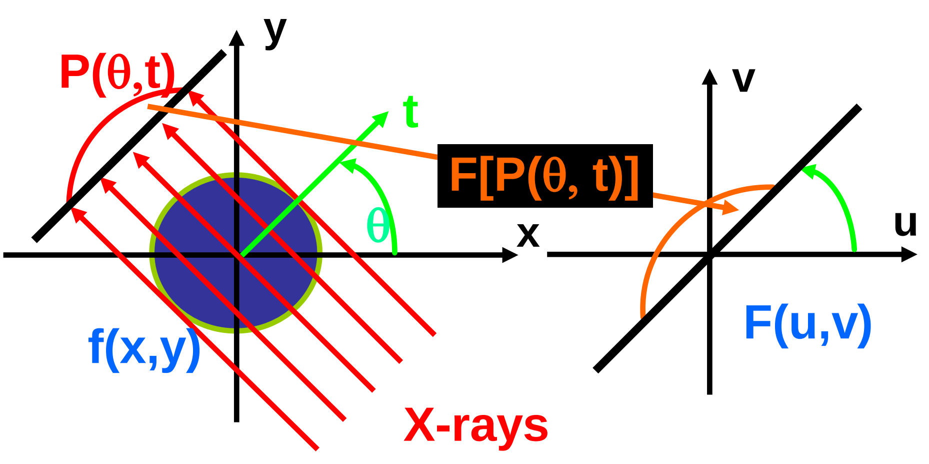


1. Convolve projections with a filter
2. Backproject filtered projections

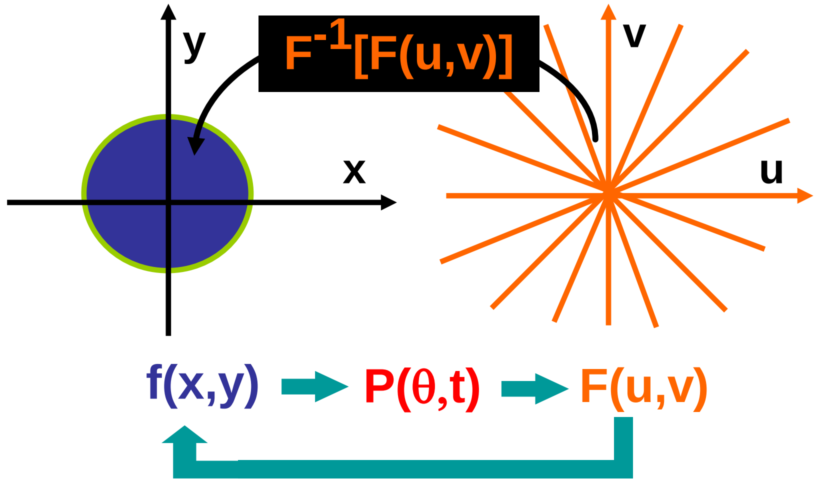
Filtered Back-projection



Fourier Slice Theorem



From Projections to Image



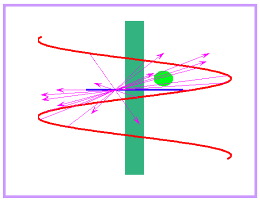
Algorithms for CT Reconstruction

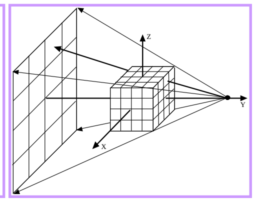
Conventional CT
- **Matrix Inversion**
- **Iterative Method**
- **Back Projection**
- **Analytical Method**
  2D Fourier Transformation
  Filtered Back-projection
  etc.

Spiral CT
- **3D Radon Transform**
- **etc.**

Corn Beam CT

3D Reconstruction



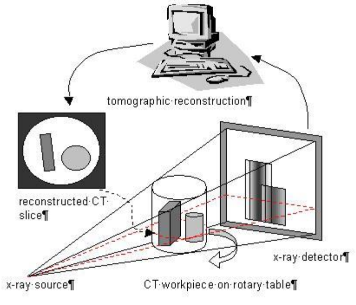


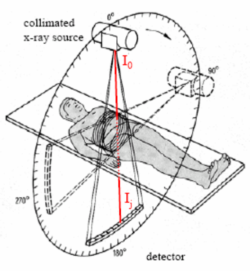
- **3D Filtered Back-projection:** based on the Feldkamp algorithm (1984). Rays are filtered and back-projected to each voxel in 3D space.
- **Nutating Slice Algorithms:** Basic idea: reconstruct tilted image planes adapted to the spiral path so that rays are close to the image plane.

Solution: Adaptive Multiple Plane Reconstruction (AMPR)

Basic Requirements of CT Image Acquisition

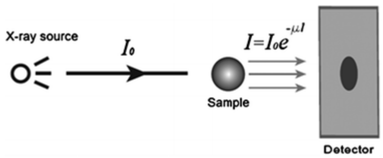
- CT image acquisition: generally 360^0 (or 180^0) direction, one tomographic image is reconstructed from x-ray projection data acquired at various angles.
- During the Scan, (1). object should be included in every projection data, (2). object have to not move.





X-ray Absorption Through the Object

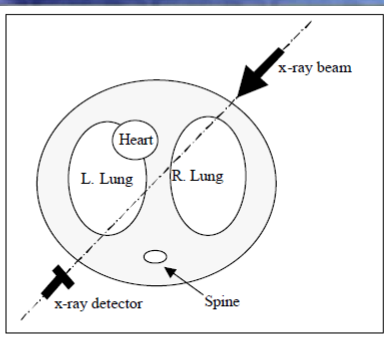
The X-rays propagate through the sample where some of the x-ray photons are absorbed, and others are transmitted to the detector. The general form of X-ray attenuation is:

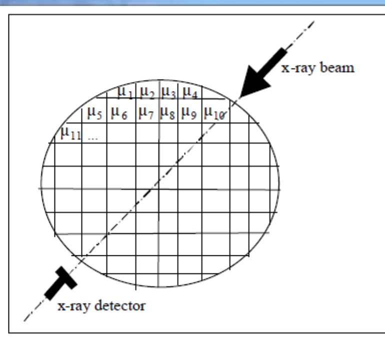


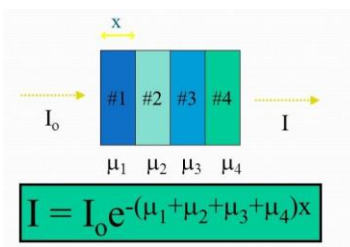
$$I_x = I_o \exp^{(-\mu x)}$$

Where:
I_0 = X-ray intensity before reaching object
I_1 = X-ray intensity after passing through object
e = the exponential coefficient (2.7182818……….)
μ = the x-ray attenuation coefficient
t = the thickness of the absorbing material, in chosen distance units e.g. mm

Attenuation Coefficient







$$\mu_{tot} = \mu_1+\mu_2+\mu_3+\mu_4$$

Illustrative example of Beer-Lambert Law, summed across multiple voxels

First Whole Body CT

- The first whole body computed tomography pictured at the Smithsonian Institution's National Museum of American History.
- This is the first model built by Dr. Robert S. Ledley and used clinically at Georgetown University hospital 1974-1978.

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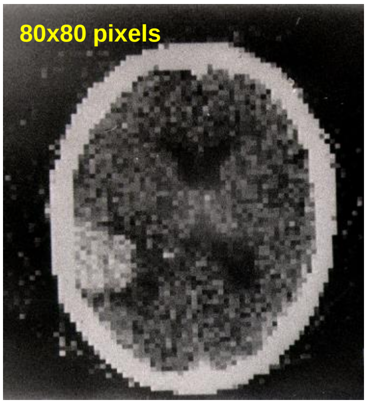
Progress in Image Quality



Siemens SIRETOM
(1974)

Siemens SOMATOM Plus 4 UFC
(1996)

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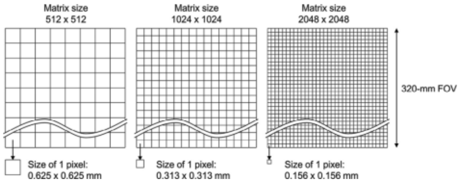
CT Image Matrix

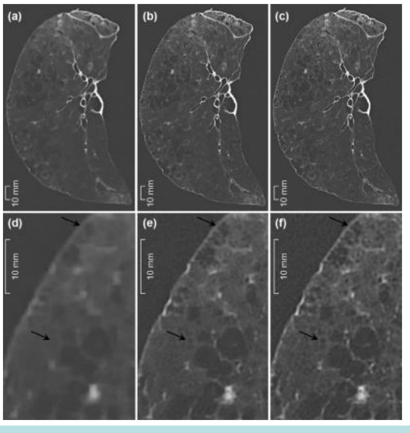
"A CT image is composed of a square image matrix that ranges in size from **256 X 256** to **1024 X 1024, 2046X2048** (Ultra high-resolution CT) picture elements or pixels.
Since a CT section has a finite thickness, each pixel actually represents a small volume element, or voxel.
The size of this voxel depends on the matrix size, the selected field of view (FOV), and the section thickness."

(Prokop and Galanski, 2003)



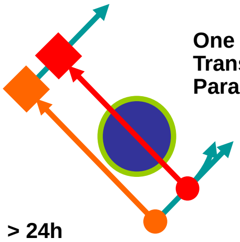
The relationship between matrix size and the size of 1 pixel. When the field of view is 320 mm, the theoretical size of 1 pixel in an image matrix is 0.625 mm for 512 × 512; 0.313 mm for 1024 × 1024; and 0.156 mm for 2048 × 2048.

CT Image Matrix



Ultra-high-resolution computed tomography images of a cadaveric lung with emphysema using 512 × 512 (a), 1024 × 1024 (b), and 2048 × 2048 (c) matrix sizes and the corresponding images that were magnified fourfold (d), (e), and (f), respectively. In the 2048 × 2048 matrix, the margin of small emphysematous lesions was depicted more clearly compared to that of the 1024 × 1024 matrix (arrows); the 512 × 512 matrix looked the most blurred and had poor quality Image noise was visually reduced in the following order: 512 × 512, 1024 × 1024, and 2048 × 2048.

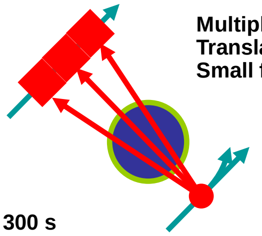
Developments of CT



One detector
Translation-rotation
Parallel-pencil beam

> 24h

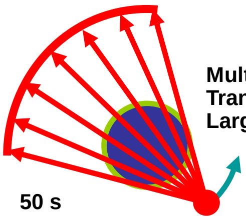
1st Generation (1970)



Multiple detectors
Translation-rotation
Small fan-beam

300 s

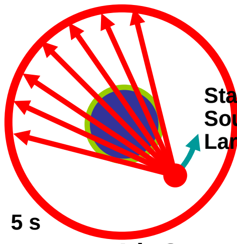
2nd Generation (1972)



Multiple detectors
Translation-rotation
Large fan-beam

50 s

3rd Generation (1976)



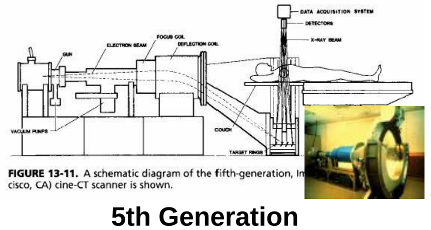
Stationary detector ring
Source-rotation
Large fan-beam

5 s

4th Generation (1978)

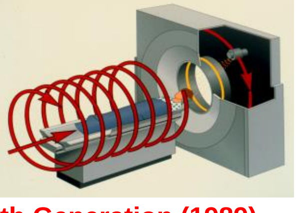
Developments of CT

Electron Beam Computed Tomography (EBCT)



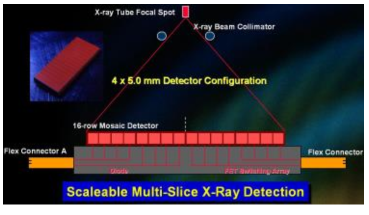
5th Generation

Spiral (Helical) CT



6th Generation (1989)

Multi-detector CT

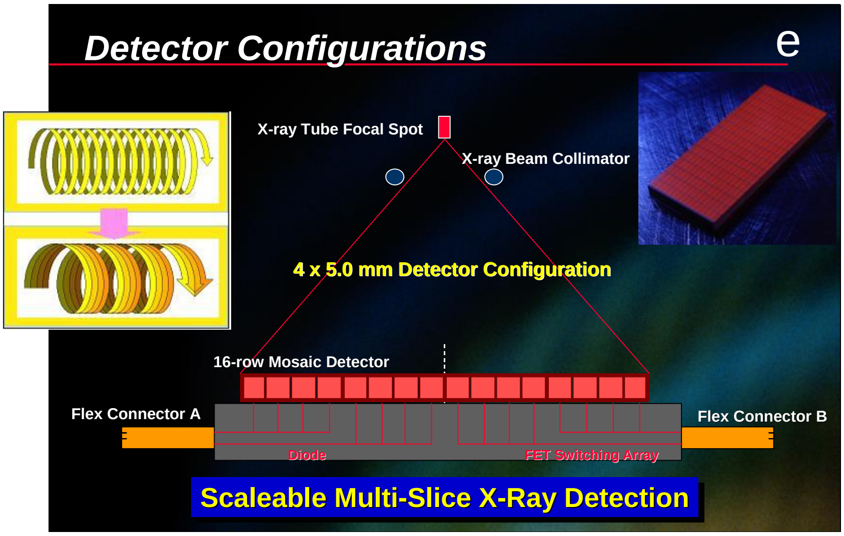


7th Generation (1998)

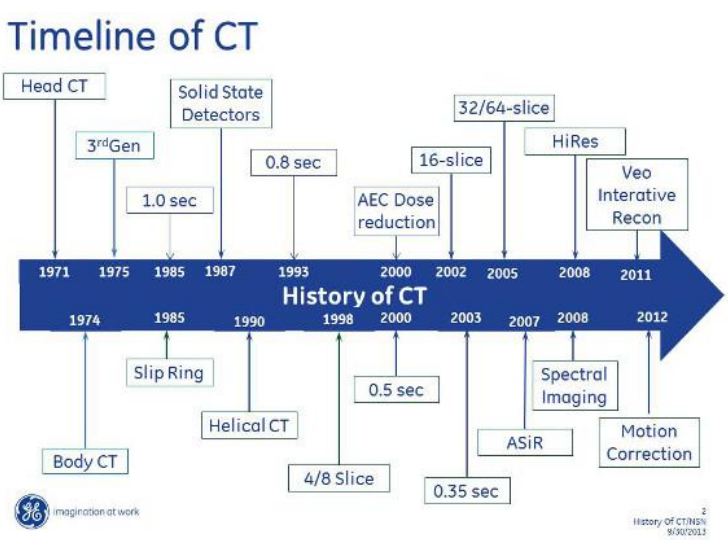
PET- CT

(2000)

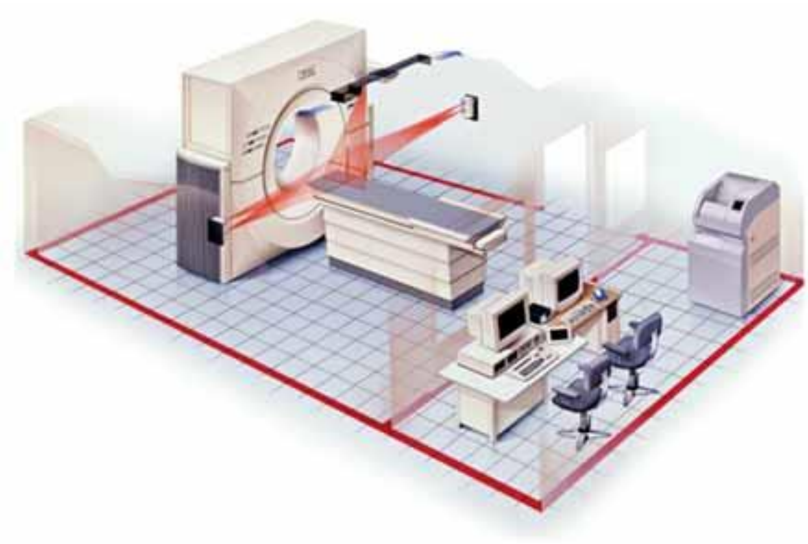
Multi-detector CT (MDCT)



Timeline of CT



CT Room



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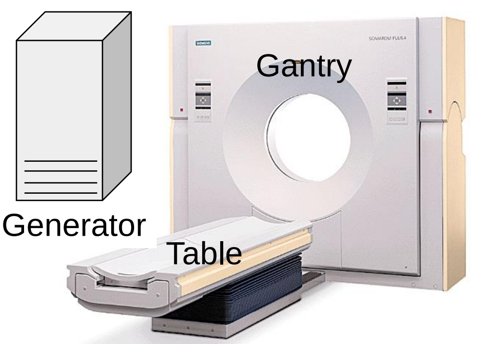
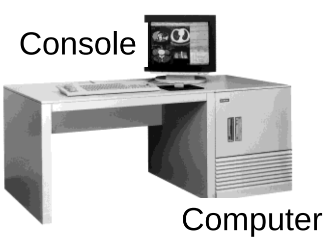
CT Components

From the outside...

- Gantry
- Table
- Generator
- Console
- Computer





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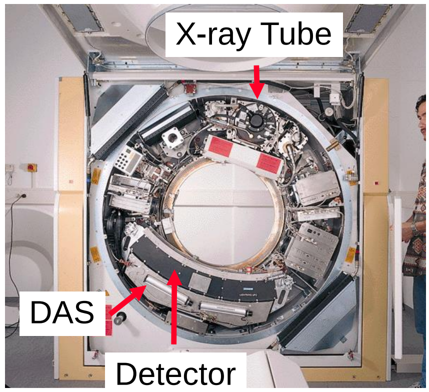
CT Gantry

From the inside

- X-ray Tube
- Detector
- DAS*

Data Acquisition System



How Dose CT Work?

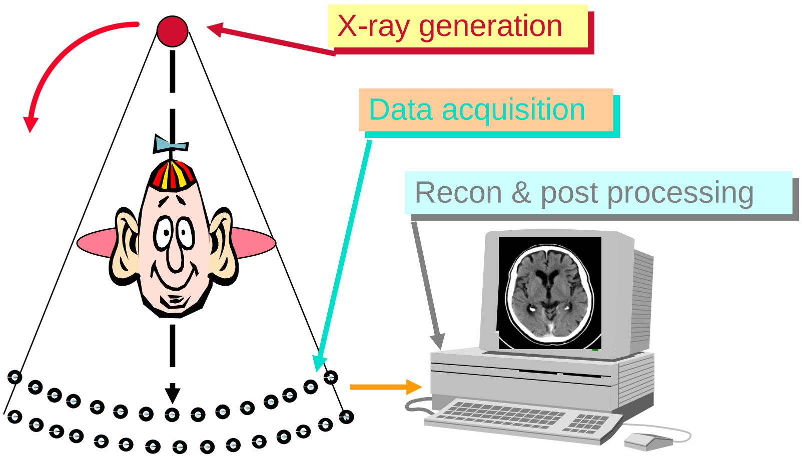
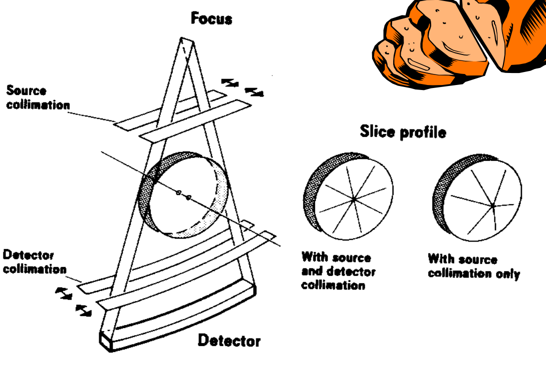


Image Generation – "Slice"

- X-rays pass through a collimator, therefore only penetrating an axial layer of the object, called a "slice"



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CT Image Generation

- The numerical matrix is converted into a black and white image in a corresponding gray scale.

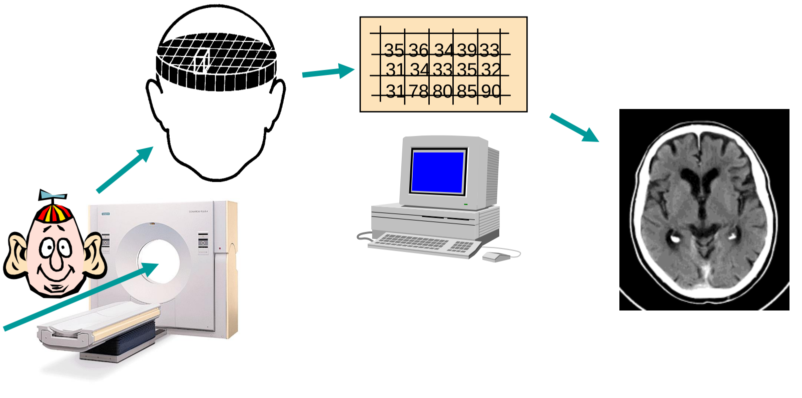

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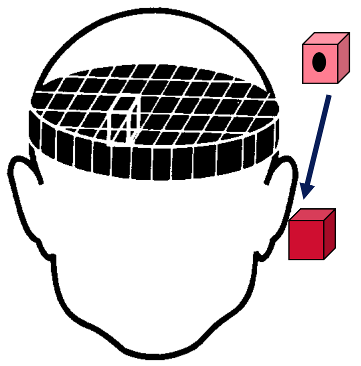
Image Generation – "Voxel"

- The slice is artificially divided into small volume elements called "voxels" with a square base, inside which the attenuation is measured as a constant value.
- And in plane, the picture elements are called "pixels"



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X-ray Detection in CT

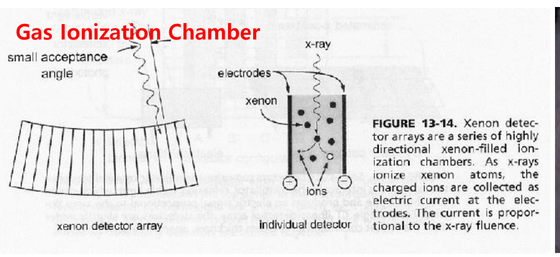

FIGURE 13-14. Xenon detector arrays are a series of highly directional xenon-filled ionization chambers. As x-rays ionize xenon atoms, the charged ions are collected as electric current at the electrodes. The current is proportional to the x-ray fluence.

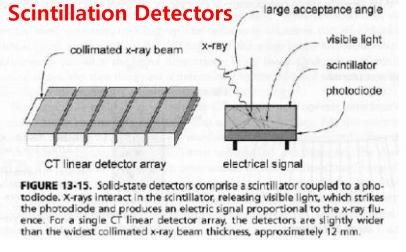

FIGURE 13-15. Solid-state detectors comprise a scintillator coupled to a photodiode. X-rays interact in the scintillator, releasing visible light, which strikes the photodiode and produces an electric signal proportional to the x-ray fluence. For a single CT linear detector array, the detectors are slightly wider than the widest collimated x-ray beam thickness, approximately 12 mm.

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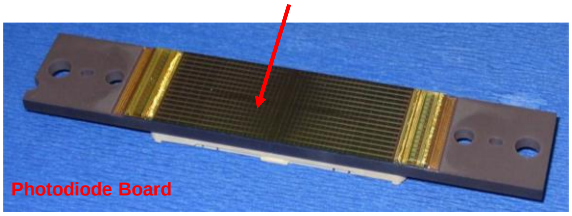
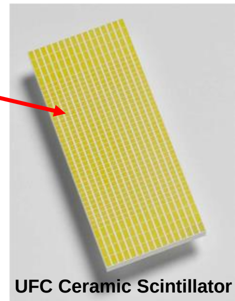
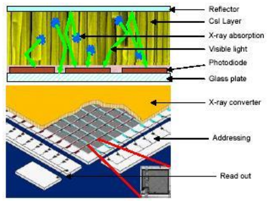
X-ray Detection in CT

Conversion X-ray → light

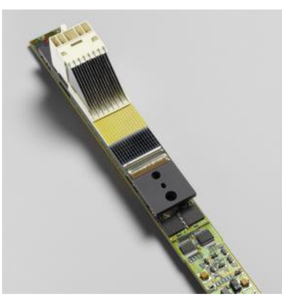
Conversion light → electrons

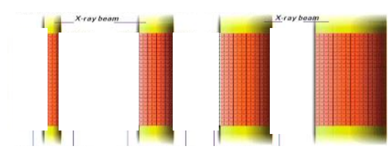


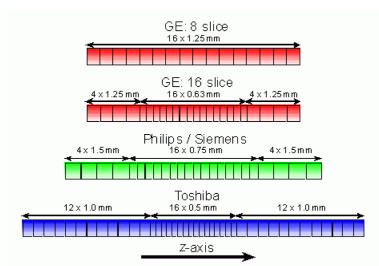




CT Detector Evolutions





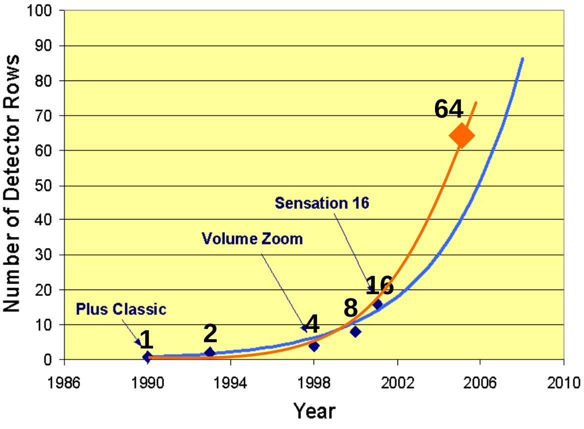


- **Early 90s:** **Spiral CT**
- **1998:** **4-slice scanners**
- **2001:** **8-16 slice scanners**
- **2003:** **32-64 slice scanners**
- **2008:** **256 slice scanners**
- **Current:** **320 slice scanner**
- **Toshiba** **320 slice scanner(?)**

Enhancement of MDCT

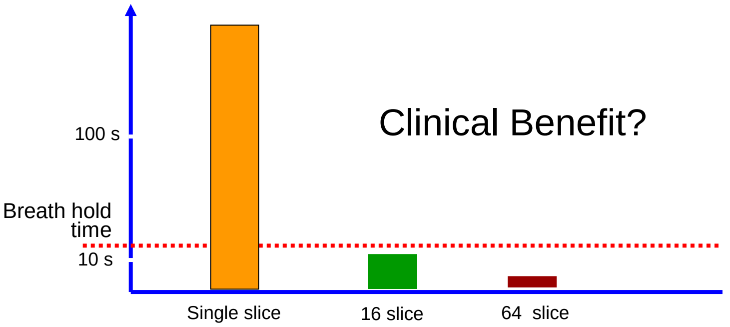
Moore's Law Valid for CT … ?

… Doubling of Slices/Rotation every ~~2.5~~ 1.5 Years



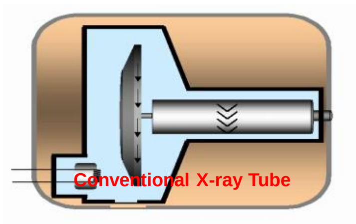
Further Development of MDCT

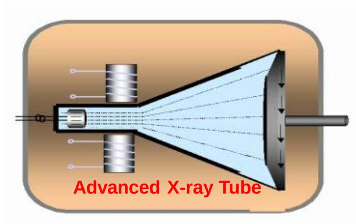
- Single slice scanner: 1 x 1 mm, 0.75 s rotation, pitch 1.5
  350 mm (entire thorax) in 177 s
- 16 slice scanner: 16 x 0.75 mm, 0.5 s rotation, pitch 1.5
  350 mm (entire thorax) in 10 s
- 64 slice scanner: 64 x 0.75 mm, 0.5 s rotation, pitch 1.5
  350 mm (entire thorax) in 2.6 s
- 128 slice scanner: 128 x 0.75 mm, 0.5 s rotation, pitch 1.5
  350 mm (entire thorax) in 1.3 s



Challenge of Tube Technology

- A modern CT tube has to sustain 60 – 80 kW for up to 20 s on a focal spot as small as 1.3x10 mm.
- With ever increasing gantry speeds, there is an extremely high mechanical load on the tube bearings.





- Direct oil cooling of the anode enables extremely high cooling rate of 5.0 MHU/min and compact design.
- Almost no anode heat storage capacity.

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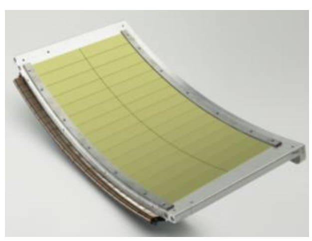
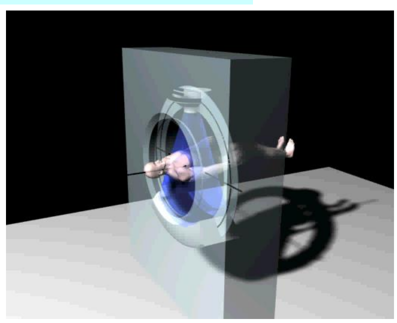
Future Trends in MDCT

Volume CT with Large Area Detectors

- Organ coverage in one shot ?
- Imaging with micro-resolution ?
- Dynamic 4D contrast studies ?
- 3D Interventions ?





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Future Trends in MDCT

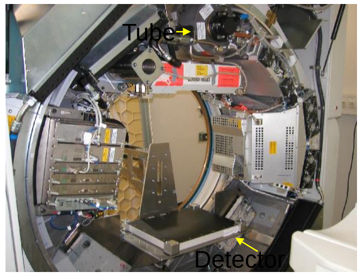
Volume CT with Large Area Detectors

Research approach: CT with flat panel detectors.

- CsI-aSi detectors known from conventional radiography.
- Excellent spatial resolution:

  e.g. 1024x768 detector channels, 0.25 x 0.25 x 0.25 mm^3 isotropic image voxels.
- Large volume coverage:

  SFOV 25 cm (in-plane) x 18 cm (z-direction).



Siemens prototype, Sensation 4 gantry,
In collaboration with MGH, Boston

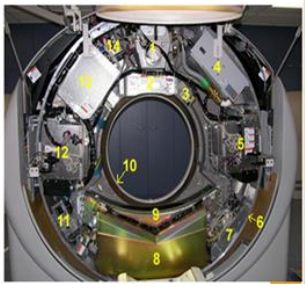
CT Table (Couch)

- ❖ CT couch (or CT table) requires the functions that are to determine the scanning position by inserting the patients to a gentry, and to control patient's movements after position settlement.
- ❖ CT couch should be strong and rigid to support weight (up to 204 kg).
- ❖ Usually made of carbon fibers due to their low absorption.
- ❖ Scan eagle range is 162 cm.

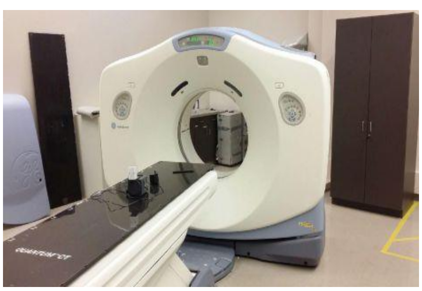
CT Gantry

CT Gantry Internal View

1. X-Ray Tube.
2. Filters, Collimator.
4. X-Ray tube heat exchanger (Oil Cooler).
5. High Voltage Generator (0-75 kV)
6. Direct Drive Gantry Motor.
7. Rotation Control Unit.
8. Data Acquisition System (DAS).
9. Detectors.
10. Slip Rings.
11. Detector Temperature Controller
12. High Voltage Generator (75-150 kV).
13. Power Unit (AC to DC).



CT Simulator used in RT



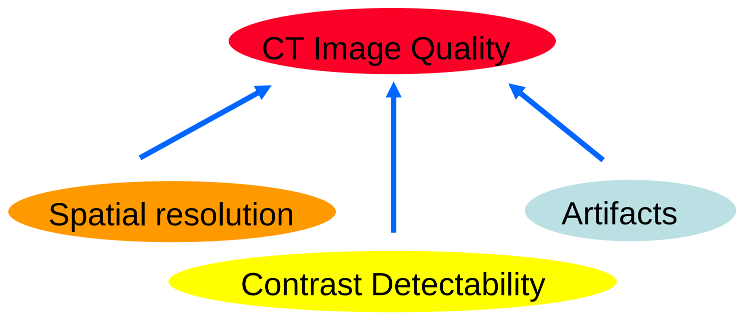
- Shape of the table is flat to get a patient image in the same condition as would be in treatment machine while a rounded table top is used in diagnostic CT for patient comport.

- Gantry bore size is bigger to accommodate situation where overall diameter of imaging volume is large due to immobilization devices and special patient postures often needed for better treatment.
  (85 cm vs. 70 cm)

CT Image Quality

- Artifact indicates what appeared in CT images, which is not existed in original clinic.
- Main reason is caused from the principal limitation of CT and the malfunction and maladjustment of CT itself.

Criteria for CT Image Quality



Influences on CT Image Quality

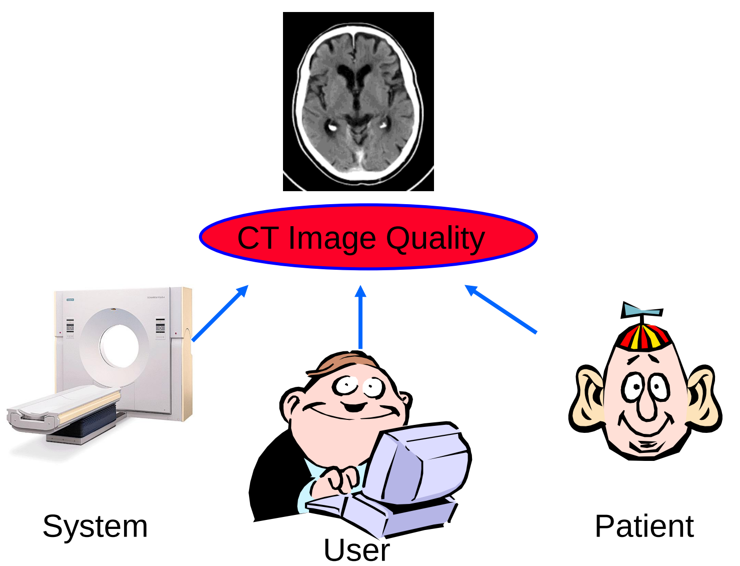
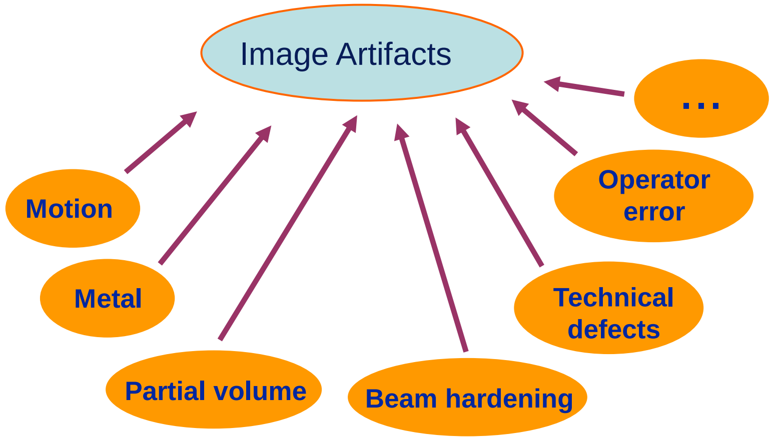


Image Artifacts - Origins



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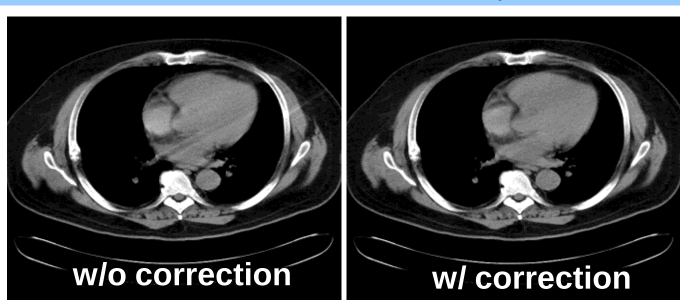
Motion Artifact & Correction

- If patients move during scan, motion artifact is occurred.
- Patient's movements should be minimized to prevent it.
- Also, It was minimized by using motion artifact correction algorithm together with control of breathing.

Motion artifacts can be compensated for by the Motion Correction Algorithm (MCA)



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Partial Volume Effect

- In complicated bone structures, the occurrence of streak artifact is more easy.
- It was appeared by "**partial volume effect**" in the case of big structural variations toward slice thickness direction.

→ (Solution: Scanning with possible thin slice thickness)

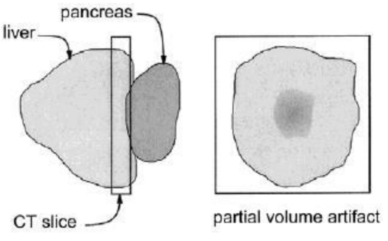
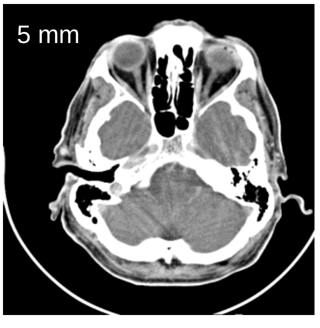


FIGURE 13-41. A partial volume artifact occurs when the computed tomographic slice interrogates a slab of tissue containing two or more different tissue types. Many partial volume artifacts are obvious (e.g., with bone), but occasionally a partial volume artifact can mimic pathologic conditions.

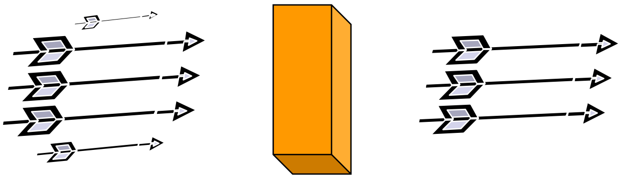
That is because the very dense structures (bones) are only partially included in the slice, resulting in high contrast errors.

Partial Volume Effect



Selecting a thinner slice prevents such artifacts from occurring, since high contrast structures are less frequently partially included, but this inherently increases the noise level, thus degrading contrast resolution.

Beam Hardening



- **Beam hardening** is the phenomenon occurring when an x-ray beam comprised of polychromatic energies passes through an object, resulting in selective attenuation of lower energy photons.
- The effect is conceptually similar to a high-pass filter, in that only higher energy photons are left to contribute to the beam and thus mean beam energy is increased ("hardened") [1].

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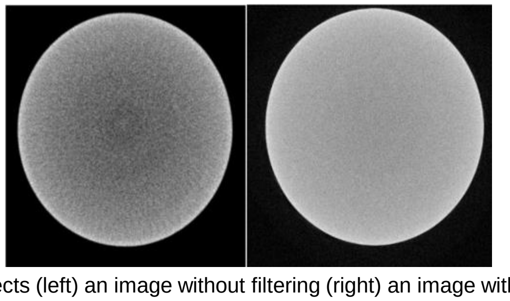
Beam Hardening

- The x-ray photons emitted from the x-ray tube do not all have the same energy.
- As they penetrate the irradiated object, the spectrum is shifted to higher energies - called "beam hardening".
- In the image, streak artifacts or the so-called "**cupping effect**" can be seen.



Beam hardening effects (left) an image without filtering (right) an image with a 1 mm thick copper filter (courtesy Nikon Metrology)

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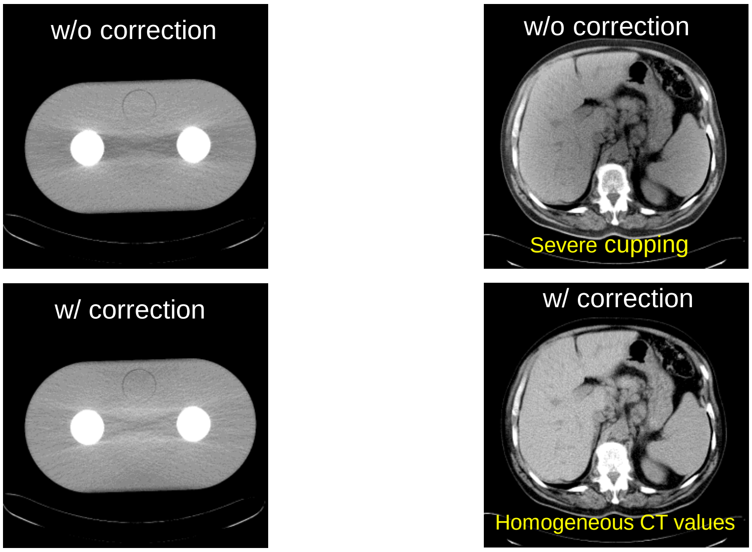
Beam Hardening & Correction



Metal Artifact

- Metal streak artifacts are caused by multiple mechanisms including beam hardening, scatter, Poisson noise, motion and edge effects in the scanned objects including hair pin, clips, and metal inside patient.

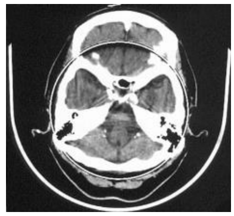
Metals, such as gold, absorb x-radiation almost completely, thus producing “radiation shadows”, leading to pronounced streak artifacts over the entire reconstructed image.

This can only be avoided via a gantry tilt that excludes the disturbing metallic objects from the slice plane.

Technical Artifacts of CT

Ring Artifact

- Ring artifacts are a CT phenomenon that occur due to miscalibration or failure of one or more detector elements in a CT scanner.
- They occur close to the isocenter of the scan and are usually visible on multiple slice at the same location.
- They are a common problem in cranial CT.



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CT Numbers/ Hounsfield Units

CT image:

- 2-D matrix of numbers.
- Each number corresponding to a spatial location in the image.

For a typical CT image:

- 512 x 512 pixels.
- 12 bits (4096 maximum) per pixel.

Other names for the numbers that correspond to the brightness of each pixel in a CT image: pixel values, gray scale values, or digital numbers, Hounsfield Units or CT numbers.

What is being measured ?

- These numbers are the average linear attenuation coefficients of the tissue.
- Scale up the linear attenuation values and normalize them to the attenuation coefficient of water.

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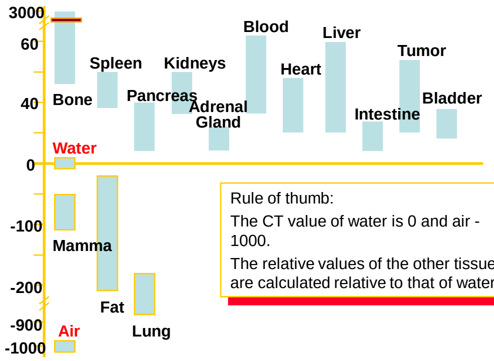
CT Numbers/ Hounsfield Units

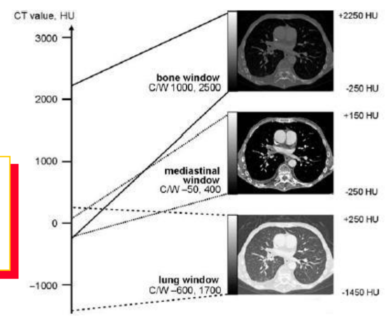
$$\text{CT number} = 1000 \times \frac{\mu_{\text{pixel}} - \mu_{water}}{\mu_{water}}$$



Rule of thumb:

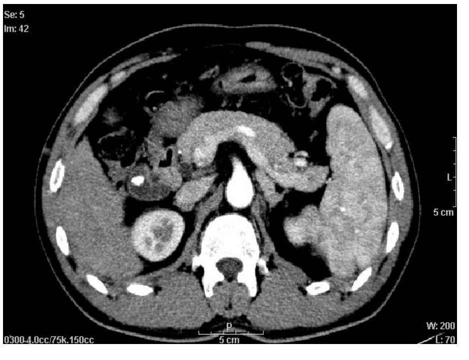
The CT value of water is 0 and air -1000.

The relative values of the other tissues are calculated relative to that of water



Normal CT Images

Axial View Images




THANK
YOU