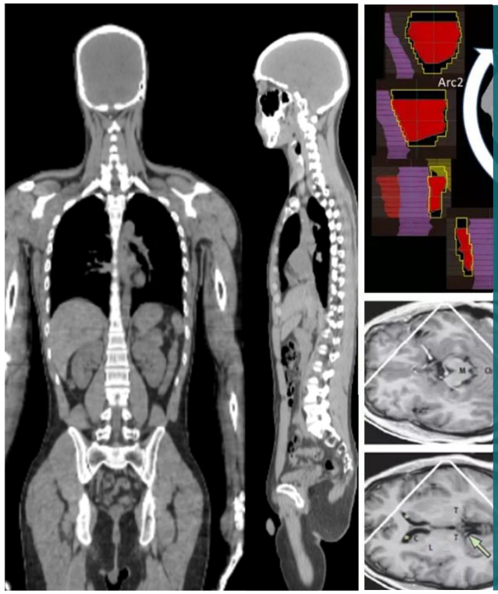
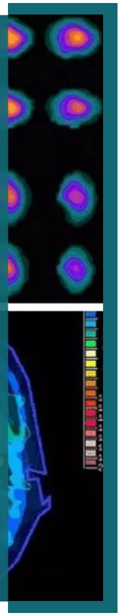




nycu

醫學影像原理與實務
電腦斷層設備與造影原理

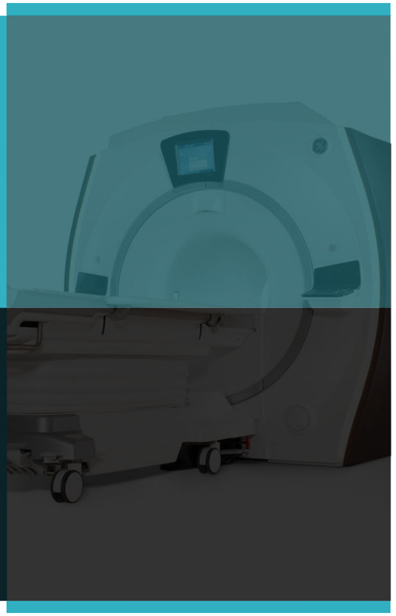
盧家鋒 教授
國立陽明交通大學
生物醫學影像暨放射科學系
alvin4016@nycu.edu.tw

本週學習目標

1. **認識X光與電腦斷層原理**
   - X光性質與X光管
   - X光與物質交互作用
   - 電腦斷層設備介紹
2. **了解電腦斷層影像重建概念**
   - 影像投影與Random轉換
   - 影像反投影與影像重建
   - 扇形投影的概念

Reference:
Fundamentals of Medical Imaging (2nd Ed.)
Chapters 2 & 3

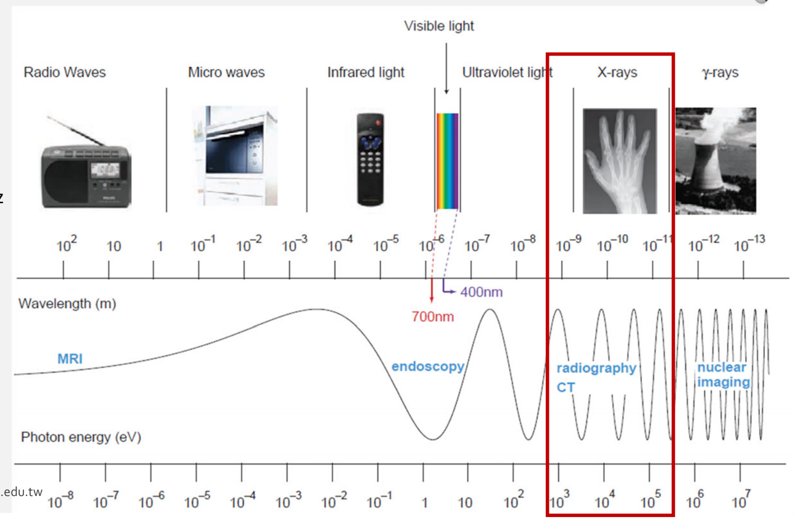
X光與電腦斷層原理



Electromagnetic Spectrum

$$E = hf = \frac{hc}{\lambda}$$

Planck's constant,
h: 4.14×10^{-15} eV/Hz
Light Speed,
c: 3.00×10^{8} m/s



Energy of X-ray
1 to 100 keV

Discovery of X-Ray

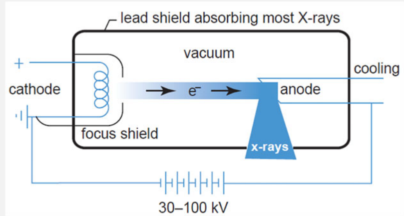
- X-rays were discovered in 1895 by **Wilhelm Conrad Röntgen**, who received the first Nobel Prize in Physics in 1901.
- "X" indicates an unknown type of radiation.

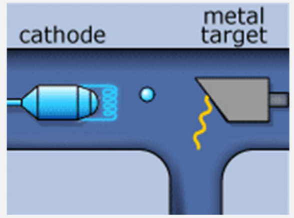
Hand with Rings: print of Wilhelm Röntgen's first "medical" X-ray, of his wife's hand, taken on 22 December 1895.

en.wikipedia.org/wiki/X-ray

X-Ray Generation

- X-rays are produced when electrons strike a metal target.
- The electrons are released from the heated filament and accelerated by a high voltage towards the metal target.
- The X-rays are produced when the electrons collide with the atoms and nuclei of the metal target.

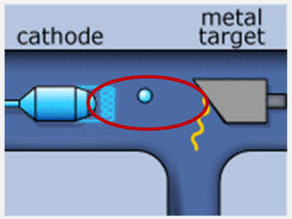




https://educationalgames.nobelprize.org/educational/physics/x-rays/how.php

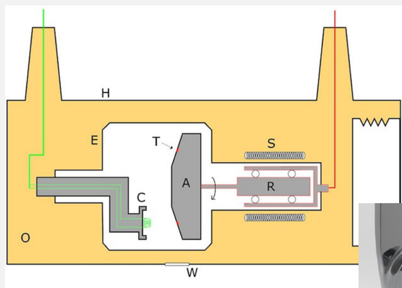
Parameters of X-Ray Source

- **Amount of emitted electrons**
  - The cathode current multiplied by the time the current is on (typically expressed in **mAs**).
  - Ranged from 1 to 100 mAs
- **Energy of emitted electrons**
  - The voltage between cathode and anode (typically expressed in **kV**).
  - For most examinations the values vary from 50 to 125 kV.
- **Total incident energy**
  - The product of the voltage, the cathode current, and the time the current is on. (typically expressed in joules, 1 J = 1 **kVmAs**)
  - 99% energy → heat and 1% → X-ray

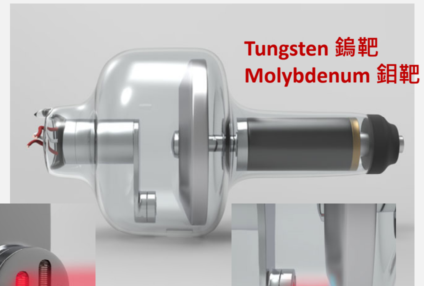


X-Ray Tube (X光管)

- filament (cathode): boils off electrons by thermionic emission
- target (anode): electrons strike to produce x-rays



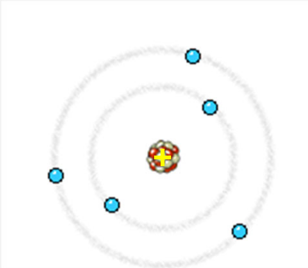
A: anode
C: cathode (and heating-coil)
E: tube envelope (evacuated)
H: tube housing
O: cooling dielectric oil
R: rotor
S: induction stator
T: anode target
W: tube window



https://radiopaedia.org/articles/x-ray-tube-1

Types of X-Ray

- **Bremsstrahlung x-ray 制動輻射/剎車輻射**
  - German for "braking“
  - Electrons lose kinetic energy as they pass through atoms in the anode.
  - Electrons are attracted to the positively charged nuclei.
  - The closer to the nucleus the electron passes, the more kinetic energy it loses.

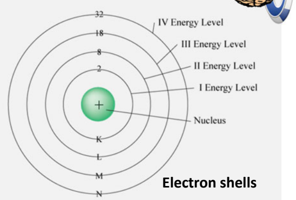


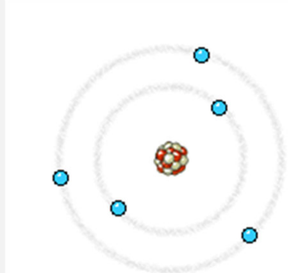
https://educationalgames.nobelprize.org/educational/physics/x-rays/what.php

Types of X-Ray

- **Characteristic x-rays 特性輻射**
  - The energy of the electrons can release an orbital electron (e.g., the K shell) in the target atom.
  - The vacancy left in the K-shell must be filled by outer-shell electrons (e.g., the L shell).
  - This process of electron transfer between shells produces "characteristic" x-rays.

$$E = E_L - E_K$$

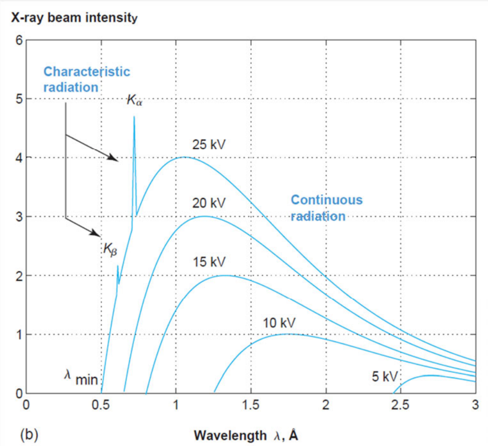




https://educationalgames.nobelprize.org/educational/physics/x-rays/what.php

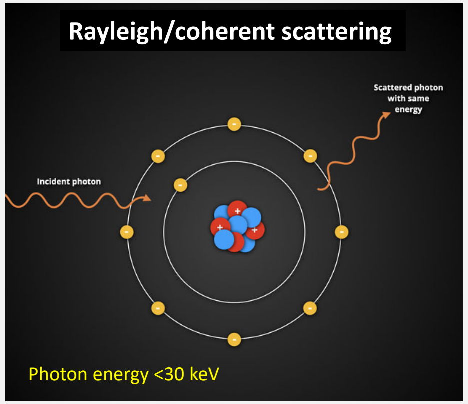
Types of X-Ray

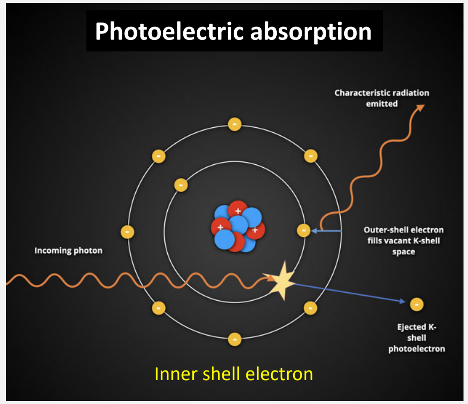
- **Bremsstrahlung x-ray**
  continuous radiation
- **Characteristic x-rays**
  If electrons have energy equal to or greater than the binding energy of the orbiting electrons in target atoms, they are likely to be ejected.



The peaks K_α and K_β are due to L-shell and M-shell drops respectively.

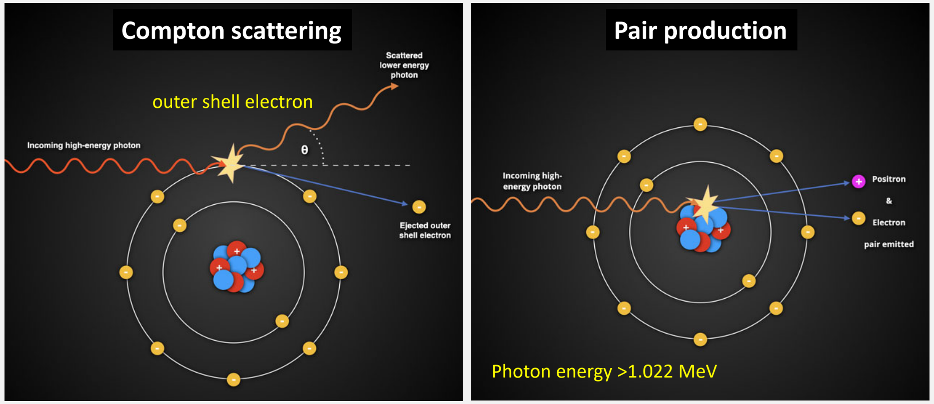
Interaction with Matter





https://radiopaedia.org/cases/types-of-photo-atom-interaction-diagrams#image-53246805

Interaction with Matter

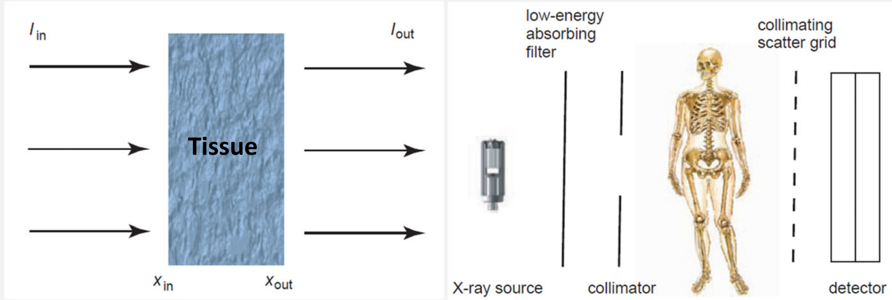


https://radiopaedia.org/cases/types-of-photo-atom-interaction-diagrams#image-53246805

Interaction with Tissue

$$I_{out} = I_{in} e^{-\int_{x_{in}}^{x_{out}} \mu(x)dx}$$

μ is linear attenuation coefficient (typically expressed in cm^{-1}).

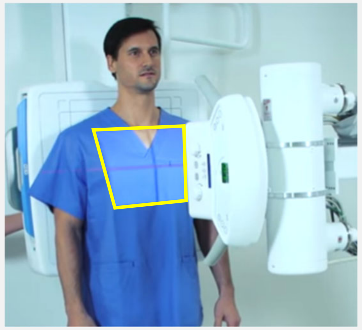


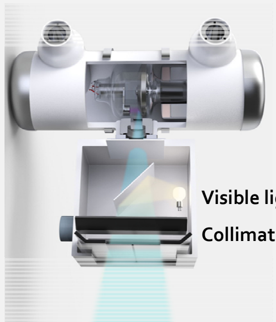
(1) film,
(2) scintillator + photodiode
(3) photon counting detector

X-Ray System

- X-ray tube, filter, collimator
- Detector/cassette/film
- Exam table
- Operating console

Field of view
照野





X-ray tube

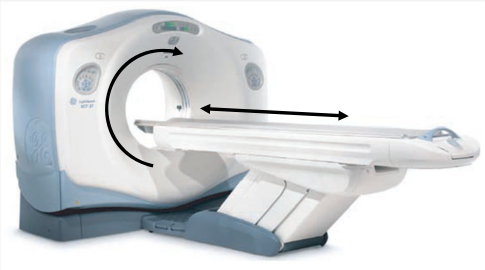
Visible light
Collimator 準直儀

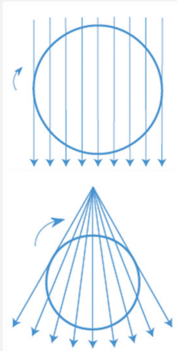
https://youtu.be/_KUmCNmpLeA?si=CAh3NA7VLJwsHnN2 https://radiopaedia.org/articles/x-ray-tube-1

https://youtu.be/l9swbAtRRbg

Computed Tomography (CT)

- CT produces cross-sectional images representing the X-ray attenuation properties of the body.



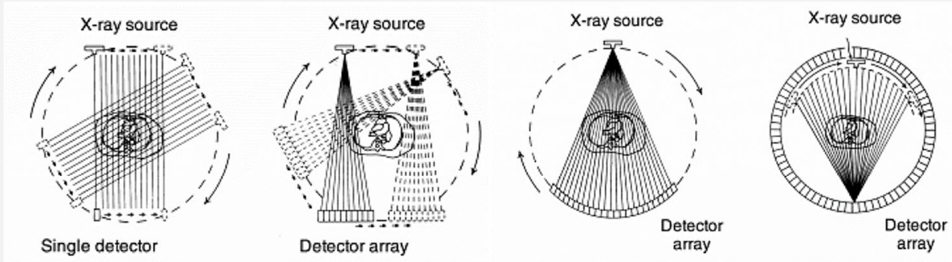


Parallel-beam geometry

Fan-beam geometry

CT Generations

1st generation | **2nd generation** | **3rd generation** | **4th generation**



Parallel/pencil beam, translate-rotate | **Fan beam, translate-rotate** | **Fan beam, rotate only** | **Fan beam, Stationary circular detector**

CT Generations

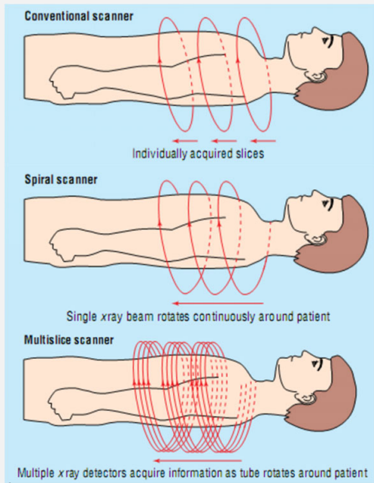
Generation	Key Features	Scanning Motion	Time per Scan	Advantages	Limitations
1st Generation (1970s)	Single X-ray tube, single detector, **pencil beam**	Translate-Rotate	4-5 minutes per slice	First medical CT, improved brain imaging	Very slow, high noise, low resolution
2nd Generation (Mid-1970s)	Multiple detectors (up to 30), **fan-shaped beam**	Translate-Rotate	20-90 seconds per slice	Faster than 1st gen, improved resolution	Still slow, limited speed for dynamic imaging
3rd Generation (Late 1970s – Present)	Curved array of detectors (hundreds), **fan beam**	Rotate-Rotate	A few seconds per slice	Faster scans, better resolution	Ring artifacts due to detector failure
4th Generation (1980s)	**Stationary detector ring**, rotating X-ray tube	Rotate-Only	<1 second per slice	No ring artifacts	Expensive due to many detectors
5th Generation (1980s – 1990s)	**Electron Beam CT (EBCT)**, no moving X-ray tube	Electron beam deflection	Milliseconds per slice	Ultra-fast, ideal for cardiac imaging	High cost, specialized use

CT Generations

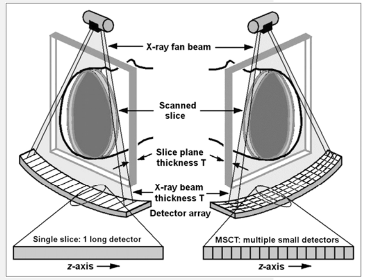
1st-5th generation

6th generation

7th generation



Fan beam, Helical/spiral Path, continuous table movement



Fan beam, Helical/spiral Path, continuous table movement

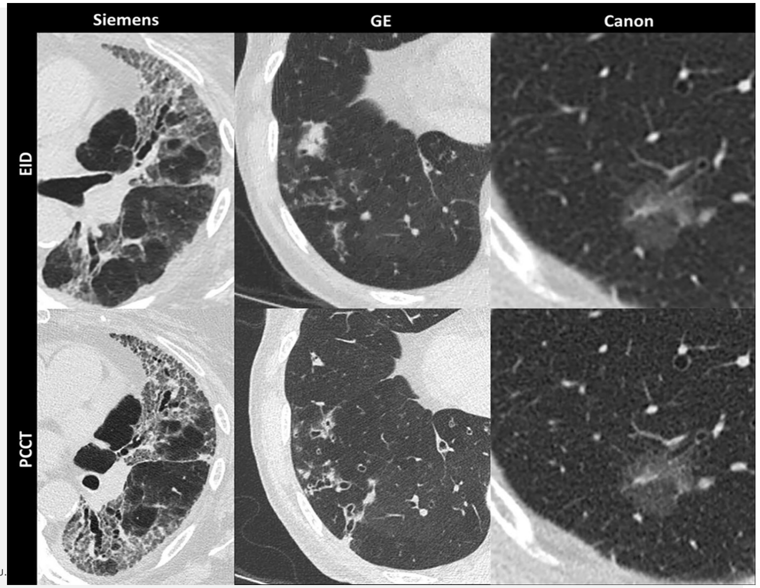
multiple detector rows (4-320 slices)

Journal of Nuclear Medicine Technology June 2008, 36 (2) 57-68.

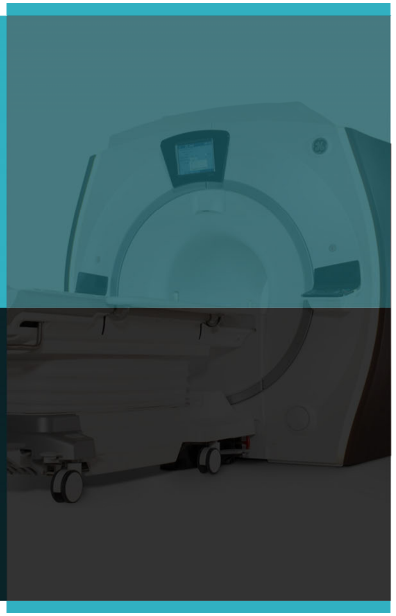
CT Generations

Generation	Key Features	Scanning Motion	Time per Scan	Advantages	Limitations
6th Generation (1990s – Present)	**Helical (Spiral) CT**, continuous table movement	**Helical Path**	Whole scan in seconds	Continuous imaging, 3D reconstruction	Requires high computing power
7th Generation (Late 1990s – Present)	**Multi-Slice CT (MSCT),** multiple detector rows (4-320 slices)	**Helical, multiple rows**	Sub-second per whole scan	High resolution, fast, larger coverage	Higher radiation dose if not optimized
8th Generation (2000s – Present)	**Dual-Source CT (DSCT)**, two X-ray tubes, two detectors	**Dual-energy helical**	<1 second for whole-body	Faster cardiac scans, material differentiation	Expensive, complex image processing
Future	**Photon-Counting CT (PCCT)**, AI-Assisted CT, Spectral CT	Varies	Real-time imaging	Lower dose, better contrast, AI enhancements	High cost, advanced processing needed

Photon Counting CT



電腦斷層影像重建



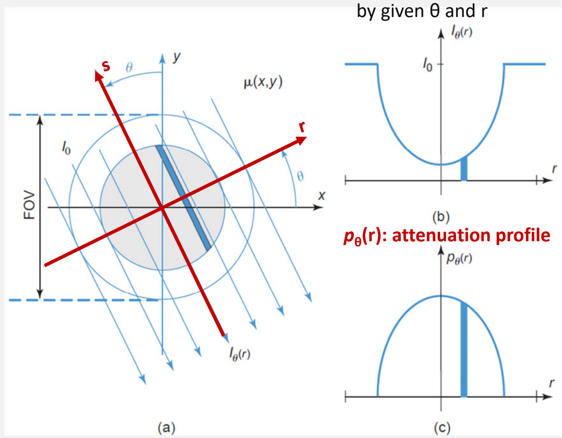
Projection in CT

- Projection of the 2D parallel-beam geometry
- X-ray beams make an angle θ with the y-axis.

$$\begin{bmatrix} r \\ s \end{bmatrix} = \begin{bmatrix} \cos\theta & \sin\theta \\ -\sin\theta & \cos\theta \end{bmatrix} \begin{bmatrix} x \\ y \end{bmatrix}$$

$$\begin{bmatrix} x \\ y \end{bmatrix} = \begin{bmatrix} \cos\theta & -\sin\theta \\ \sin\theta & \cos\theta \end{bmatrix} \begin{bmatrix} r \\ s \end{bmatrix}$$

$I_\theta(r)$: projected intensity profile by given θ and r



Projection and Random Transform

- Stacking all the attenuation profiles $p_\theta(r)$ results in a 2D dataset ***p*(r, θ)** called a **sinogram.**

$$p(r,\theta) = R\{f(x,y)\}$$
$$= \int_{-\infty}^{\infty} f(r \cdot cos\theta - s \cdot sin\theta, r \cdot sin\theta + s \cdot cos\theta) ds$$

- The transformation of a function *f*(x,y) into its sinogram *p*(r,θ) is called the **Radon transform**.

https://youtu.be/LS4C7-GE_zE?feature=shared

Projection and Random Transform

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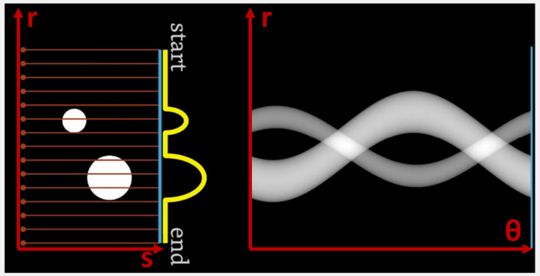

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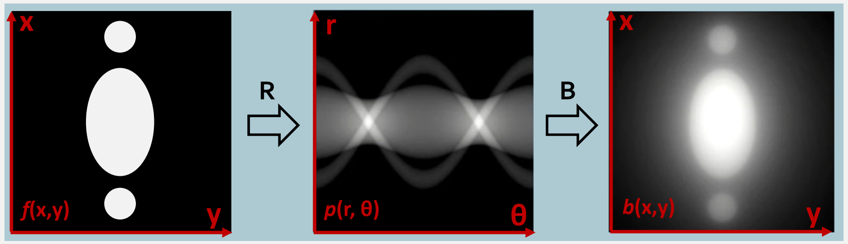
https://youtu.be/LS4C7-GE_zE?feature=shared

Backprojection

- **Backprojection** is the procedure to reconstruct the *f*(x,y) based on the signogram *p*(r, θ).

$$b(x,y) = \mathcal{B}\{p(r,\theta)\} = \int_0^{\pi} p(x \cdot cos\theta + y \cdot sin\theta, \theta) d\theta$$

任一點*f*(x,y)的數值，可以藉由把不同角度的投影量*p*(r, θ)作加總來反推

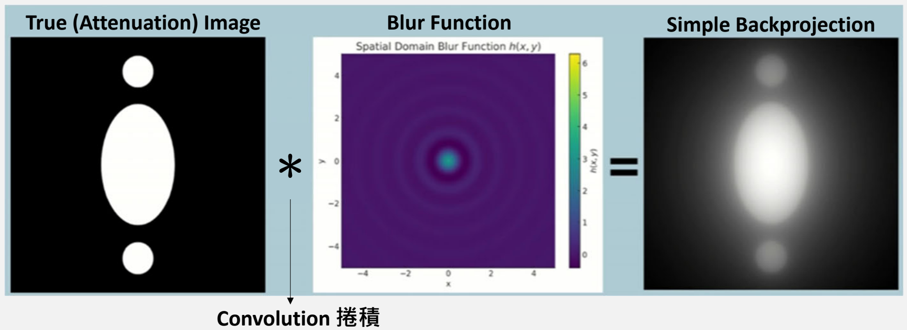


Simple backprojection creates **a blurred reconstructed image**.

https://youtu.be/ahpFGvyM0Ak?feature=shared

Effect of Blur Function

- The simple backprojection produces image similar to the true image convolved with a blur function.



https://youtu.be/3lqYMn1PYIE?feature=shared

Filtered Backprojection

- **Backprojection** is now performed on the filtered sinogram *p**(r, θ) to eliminate the “blurring” effect.

$$f(x,y)=\int_0^{\pi} p^*(x\cdot cos\theta+y\cdot sin\theta,\theta)d\theta$$

Convolution with the inverse Fourier transform of the **ramp filter |k|**.

$$p^*(r,\theta)=\int_{-\infty}^{\infty} p(r',\theta)q(r-r')dr'$$

A high-pass filter that removes low-frequency noise.

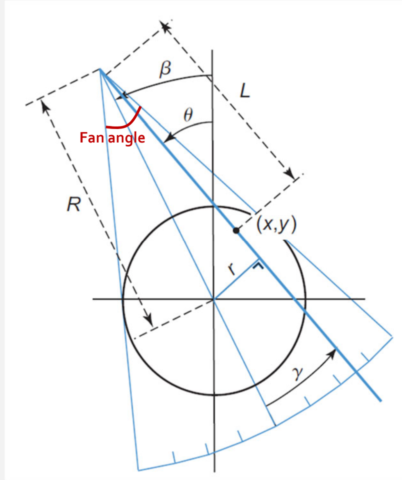
$$q(r)=\mathcal{F}^{-1}\{|k|\}=\int_{-\infty}^{\infty}|k|e^{i2\pi kr}dk$$

 https://youtu.be/3lqYMn1PYIE?feature=shared (17:48)

Fan-beam Backprojection

- A range from **0 to (π + fan-angle)** is required to include all line measurement.
- Two reconstruction approaches:
  - Rearrange data into parallel geometry with interpolation.
  - An adapted equation for filtered backprojection.

$$\theta=\beta+\gamma$$
$$r=Rsin\gamma$$



重點回顧

- X光可透過加速熱電子撞擊鎢或鉬靶來產生。
- 制動輻射為連續能譜，而特性輻射則有特定能峰。
- 診斷用的X光與CT，物質主要透過光電效應(photoelectric effect)與康普頓散射(Compton scattering)衰減X光能量。
- 不同世代CT之演進，與射源、接收器之配置、移動方式有關。
- CT造影過程中，會透過不同角度之投影來收集sinogram。
- 取得之sinogram則可透過濾波反投影(filtered backprojection)來進行影像重建。

BIRS

Department of Biomedical Imaging and Radiological Sciences, NYCU

盧家鋒 教授

國立陽明交通大學 生物醫學影像暨放射科學系

alvin4016@nycu.edu.tw

網頁: http://cflu.lab.nycu.edu.tw