# 輻射偵測與造影原理

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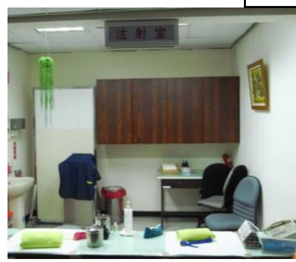
# 核子醫學的原理

- George de Hevesy，1885-1966，(1943年諾貝爾化學獎得主）
- 核子醫學檢查的基本原理是利用放射性物質所標化的藥物作為放射性追蹤劑（Radiotracer)
- 追蹤劑原理(Tracer Principle):
  - Georg Charles De Hevesy
  - 利用極少量的放射性藥物進入人體內，再以核醫儀器追蹤或造影。

# 核醫部工作流程

病患 → 櫃檯 → 注射室 ← 製藥

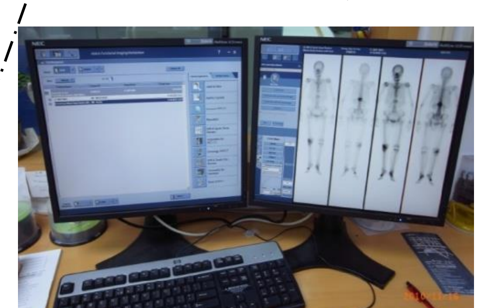

注射室 → 候診室

候診室 → 造影

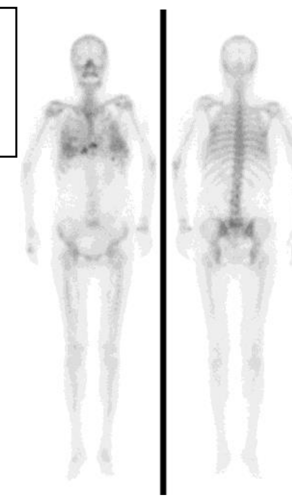
櫃檯 → 造影


造影 → 影像後處理與上傳 → 醫師影像判讀

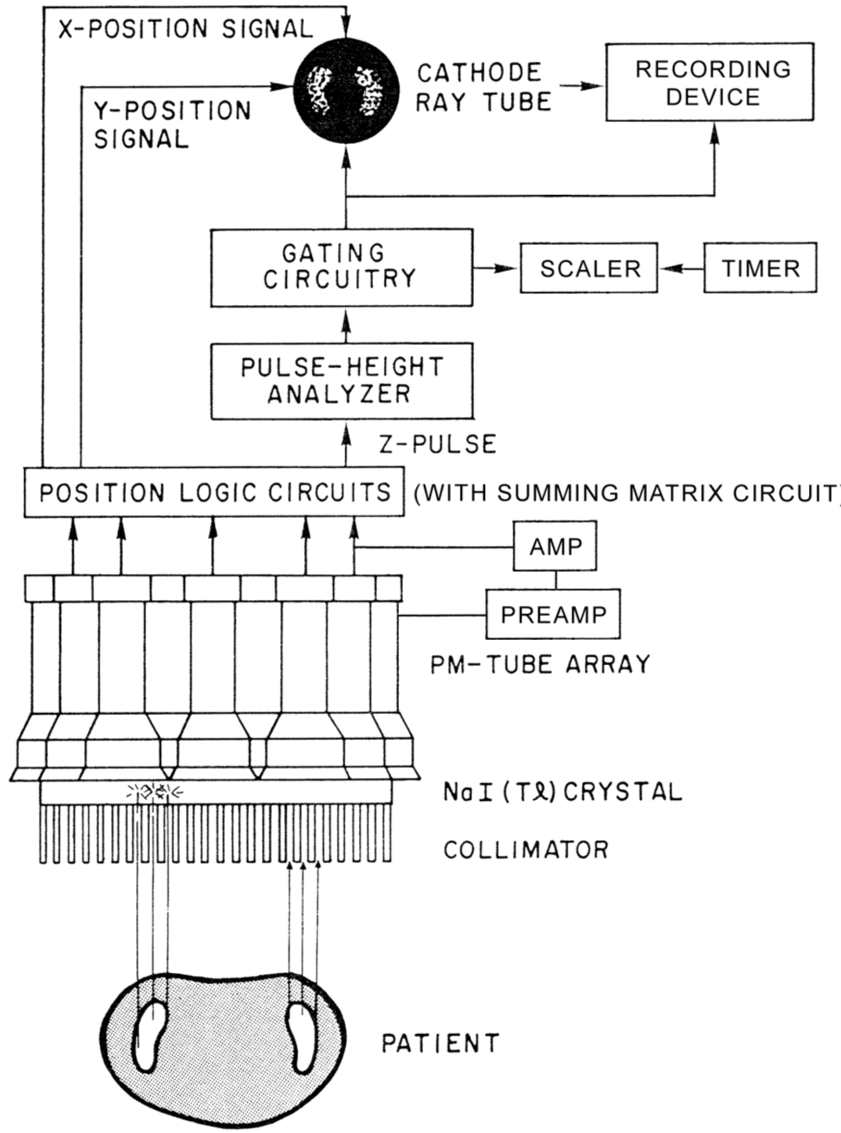
造影 → 電腦組



醫師影像判讀 → 報告



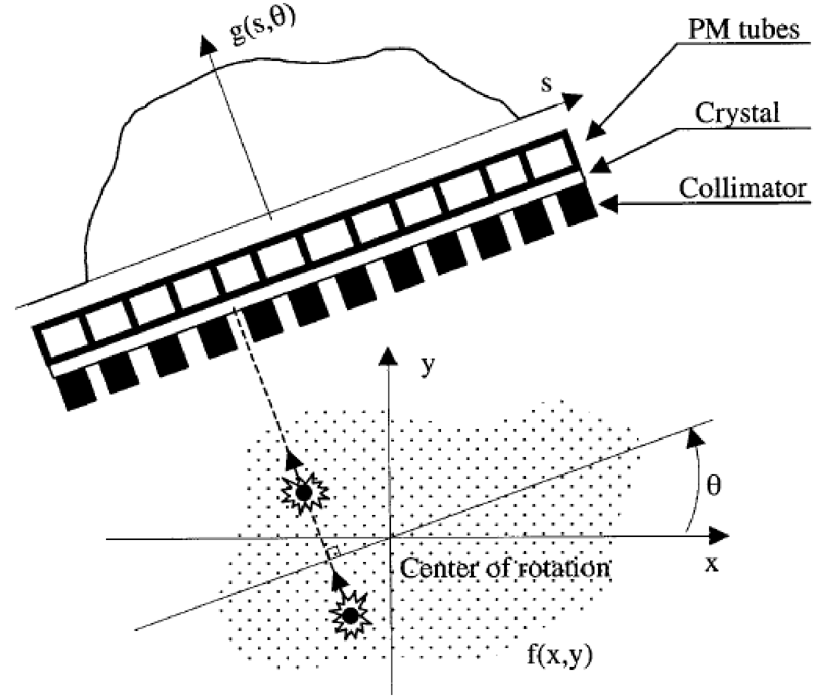
# 閃爍造影機系統組成元件



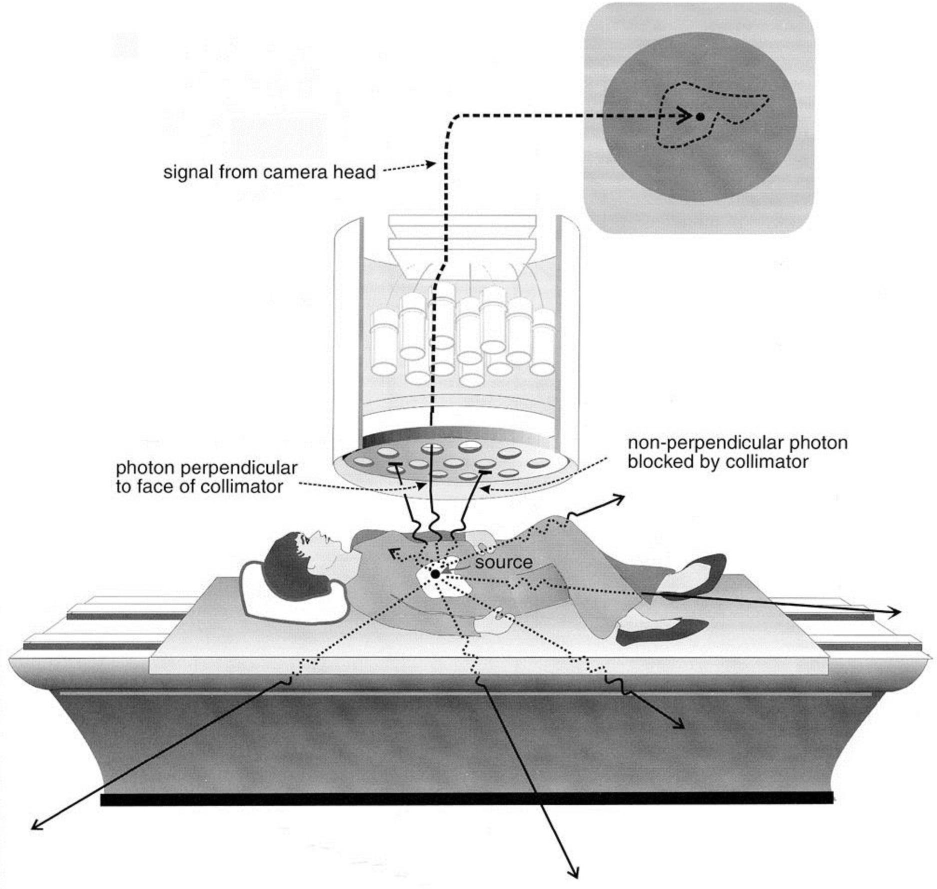

Basic principles and components of the Anger camera.

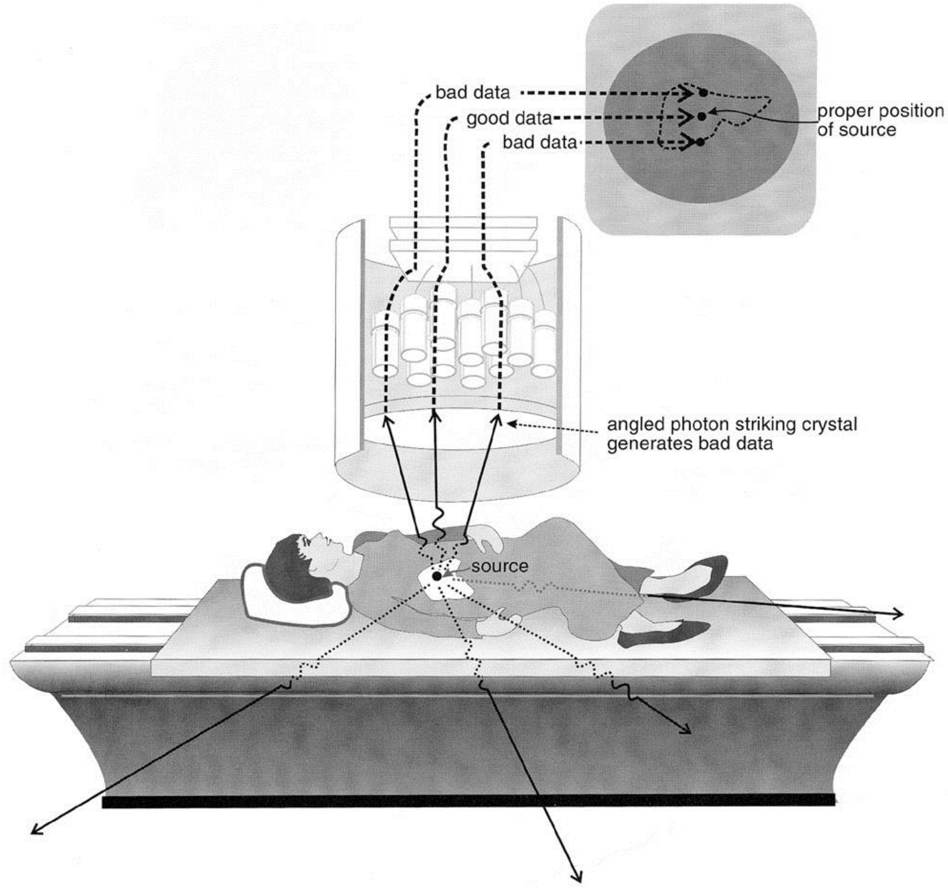
# 核子醫學成像方式

- 為放射式 (Emission)
- 光子的數目少
- 偵測單位範圍大
- 解析度低
- 為功能性影像



# 準直儀（Collimator）



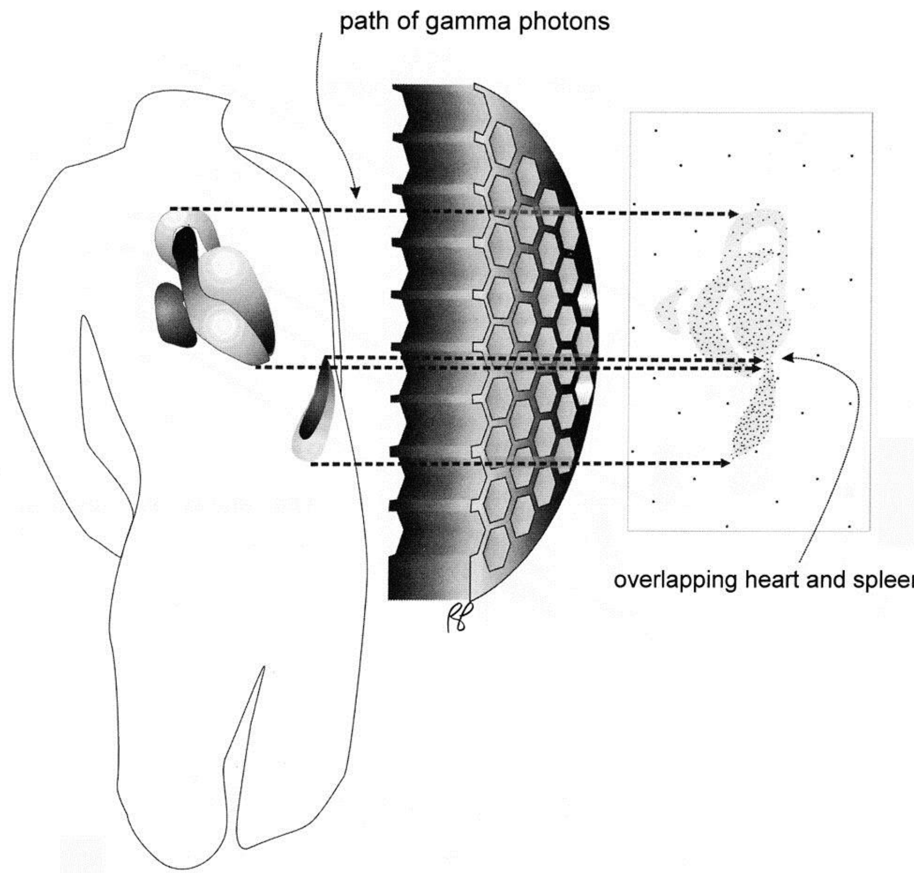

With Collimator

Without Collimator

# 平行式多孔準直儀（Parallel-Hole Collimator）

其(影像)特性為：

- 由許多孔徑大小一致的鉛或鎢製準直孔所組成，為目前臨床最普遍使用的準直儀。
- 影像大小不受射源(目標器官)到準直儀表面間距離的影響。
- 幾何效率(Geometric Efficiency)對空間距離反應平均。
- 系統解析度(System Resolution)在其表面最佳，但隨著射源(目標器官)到準直儀表面間距離的增加，準直儀解析度會迅速地下降。



# 針孔式單孔準直儀（Pinhole Collimator）

## 放大因子(magnification factor)：

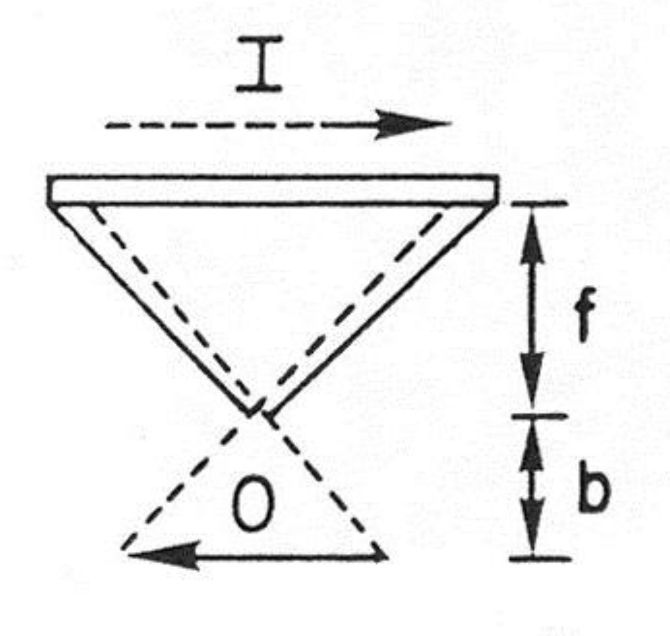
$$I/O = f/b$$

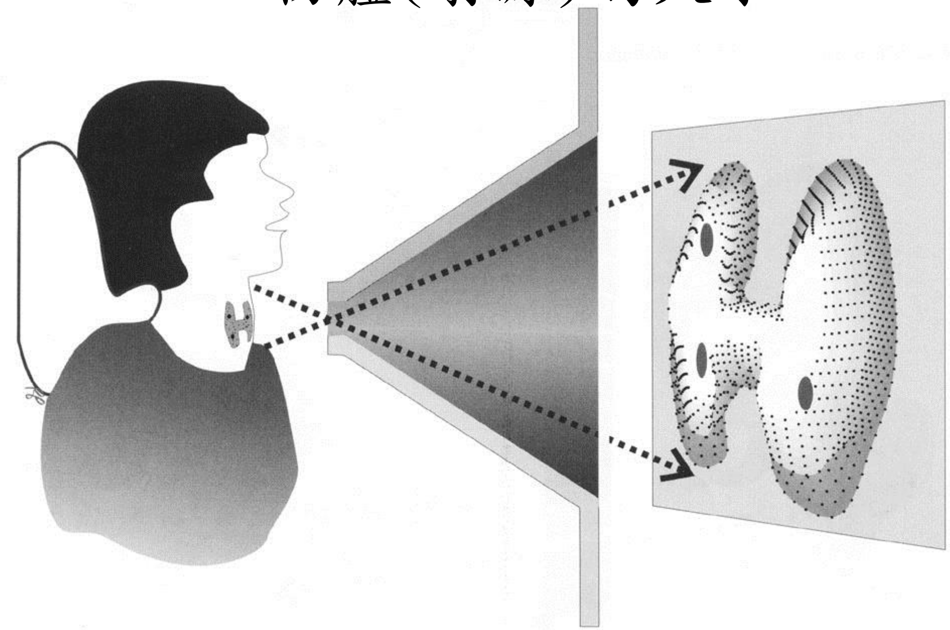
*f*: 準直儀圓錐的垂直長度

*b*: 物體(射源)至針孔的距離

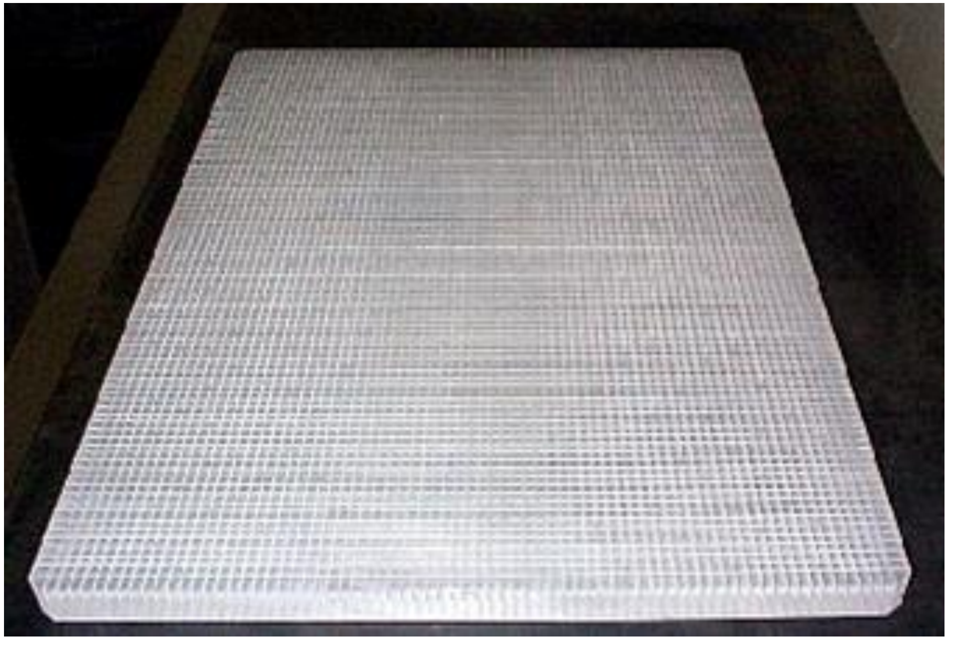
*I*: 影像的大小

*O*: 物體(射源)的大小

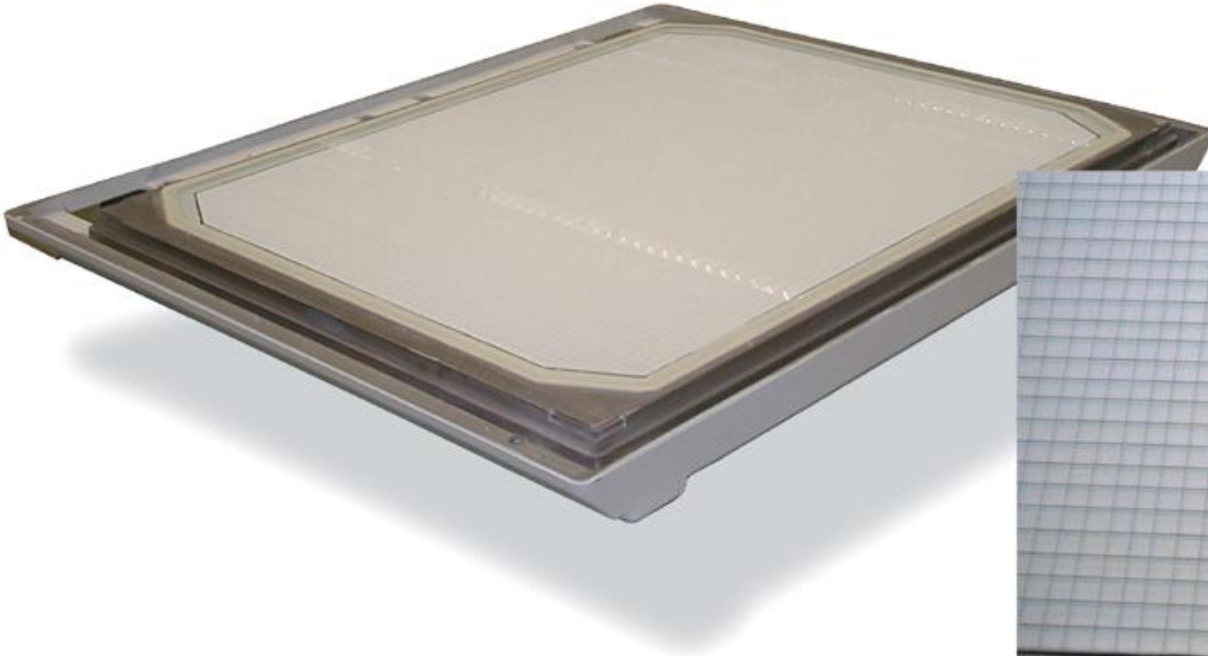


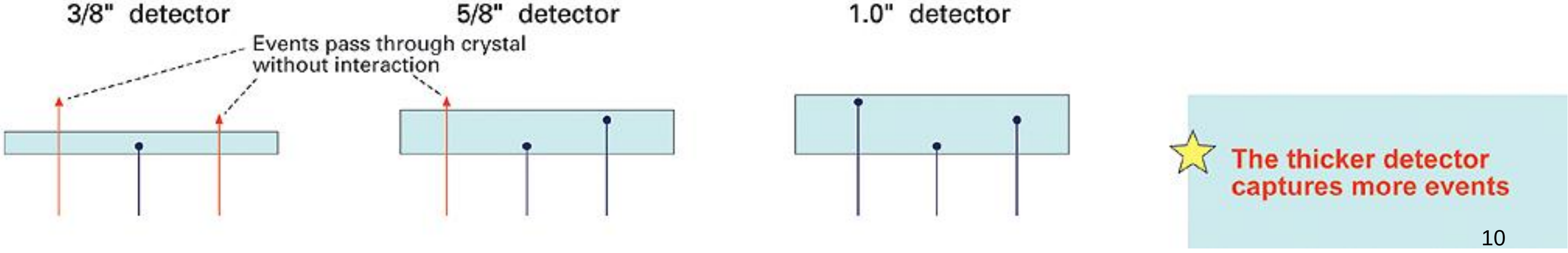



# 閃爍晶體（Scintillation Crystal）

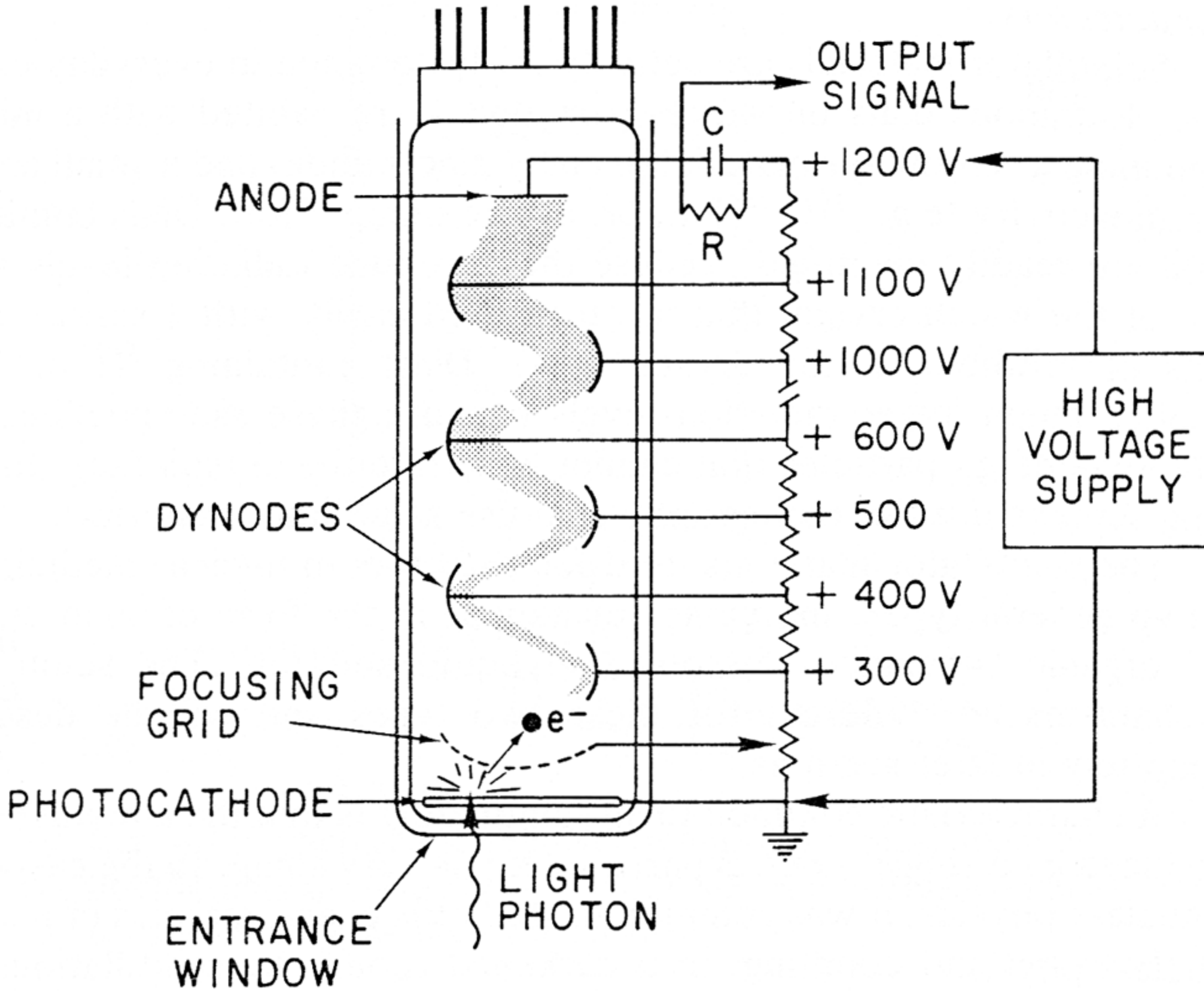


# 閃爍晶體（Scintillation Crystal）

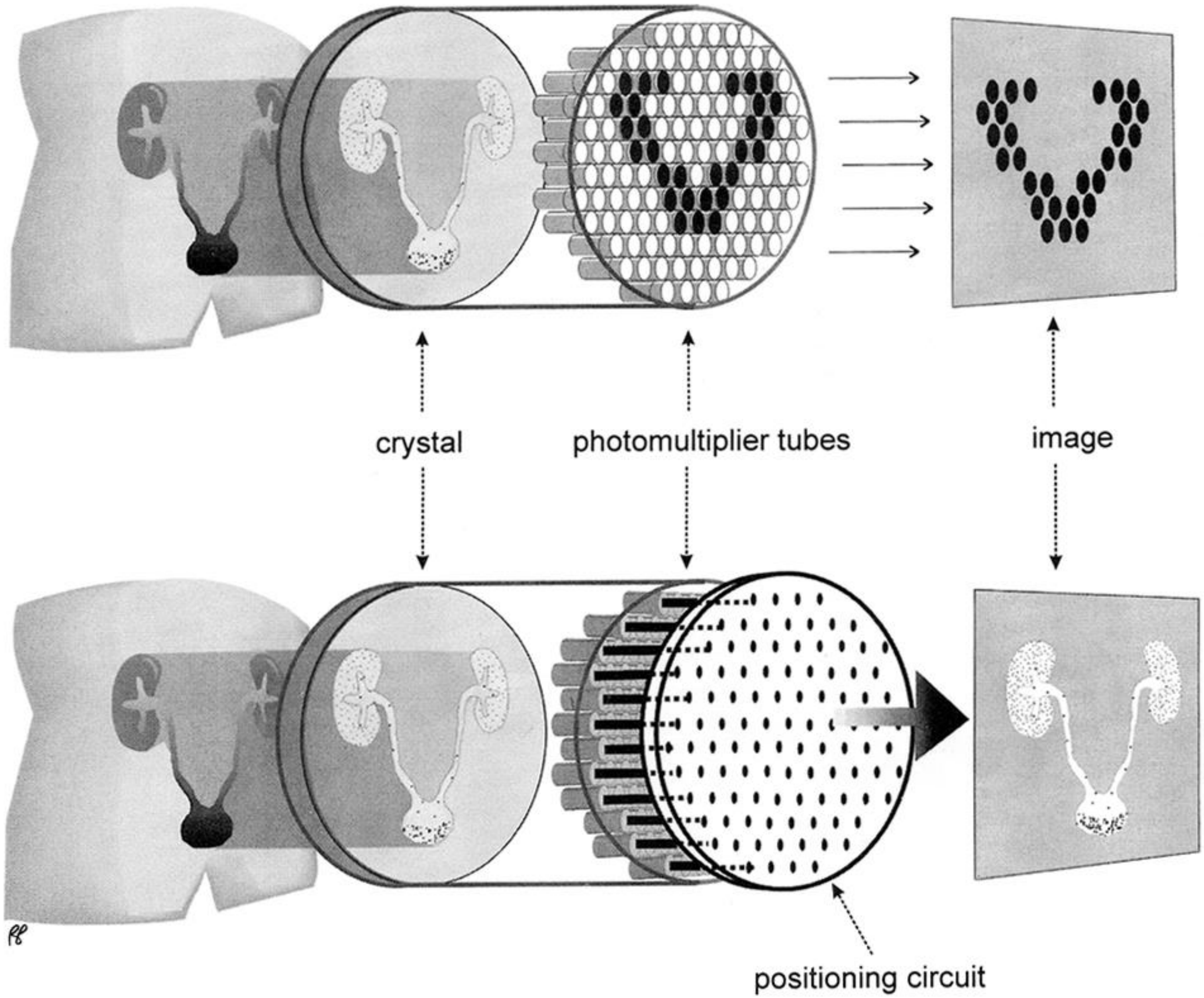




# 光電倍增管（Photomultiplier Tube）

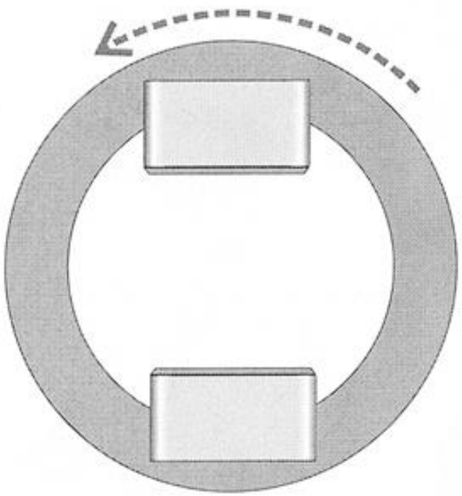


# 位置邏輯電路（Position Logic Circuit）

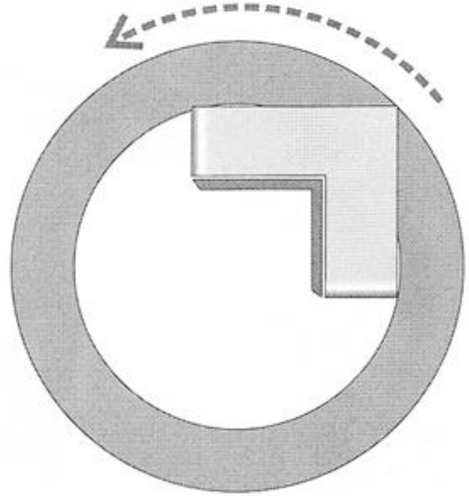



*The positioning circuit improves image resolution.*

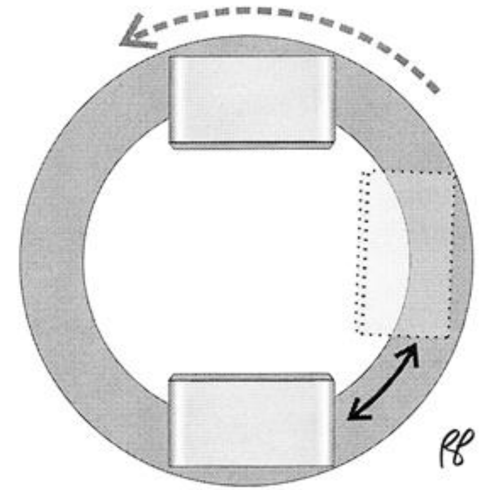
# 雙偵測頭單光子射出電腦斷層掃描儀



fixed, parallel



fixed, perpendicular



adjustable

*Two-headed SPECT camera.*

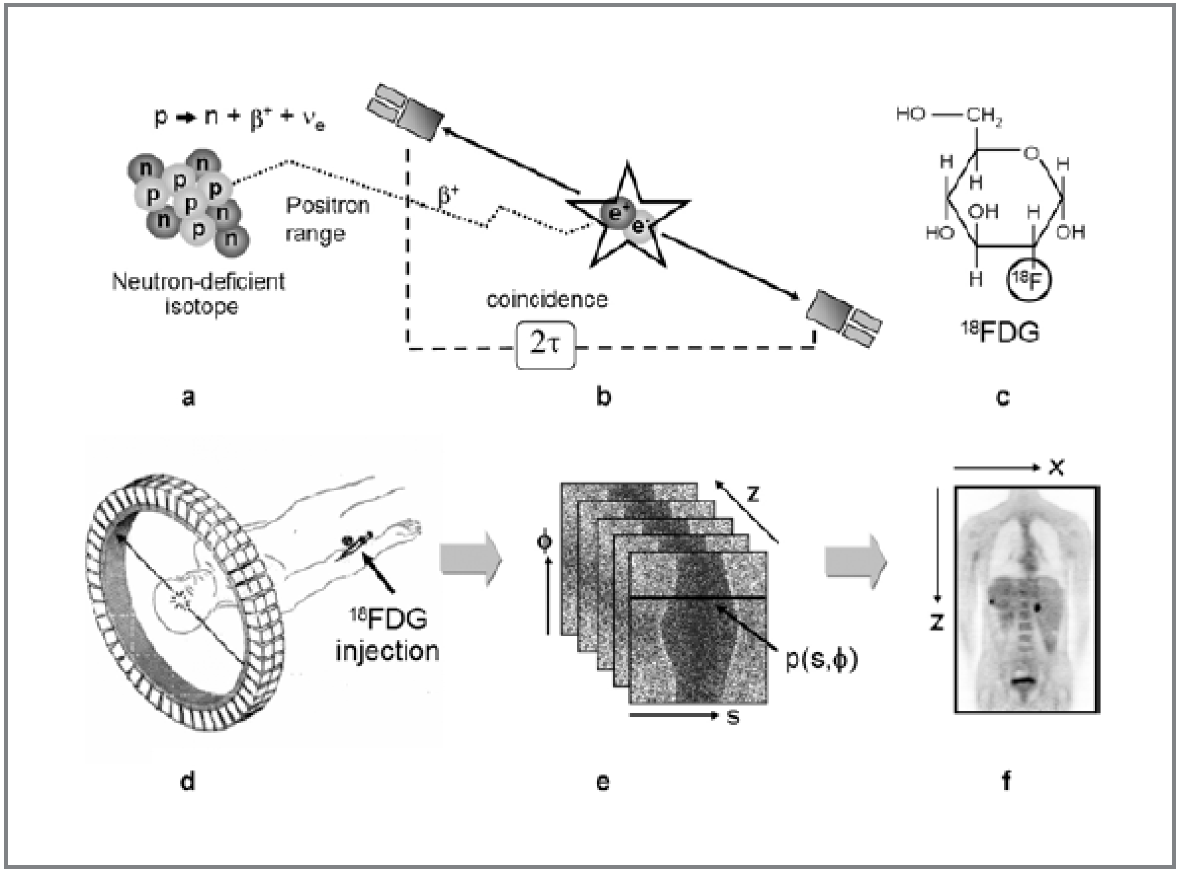
180 度斷層掃描

90 度斷層掃描

76 度斷層掃描

# PET Overview

**Figure 1.1.** The principles of PET imaging shown schematically: (**a**) the decay of a neutron-deficient, positron-emitting isotope; (**b**) the detection in coincidence of the annihilation photons within a time window of 2τ ns; (**c**) the glucose analogue deoxyglucose labeled with the positron emitter $^{18}F$ to form the radiopharmaceutical FDG; (**d**) the injection of the labeled pharmaceutical and the detection of a pair of annihilation photons in coincidence by a multiring PET camera; (**e**) the collection of the positron annihilation events into sinograms wherein each element of the sinogram contains the number of annihilations in a specific projection direction; and (**f**) a coronal section of the final, reconstructed whole-body image mapping the utilization of glucose throughout the patient.



# Positron Annihilation

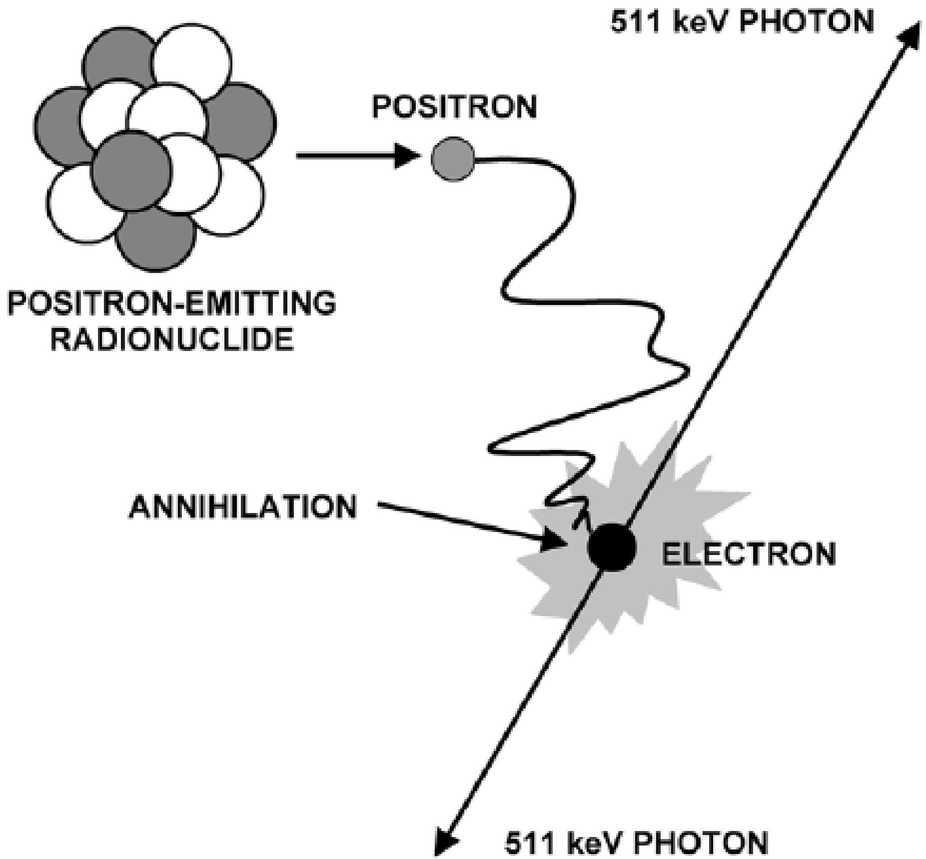
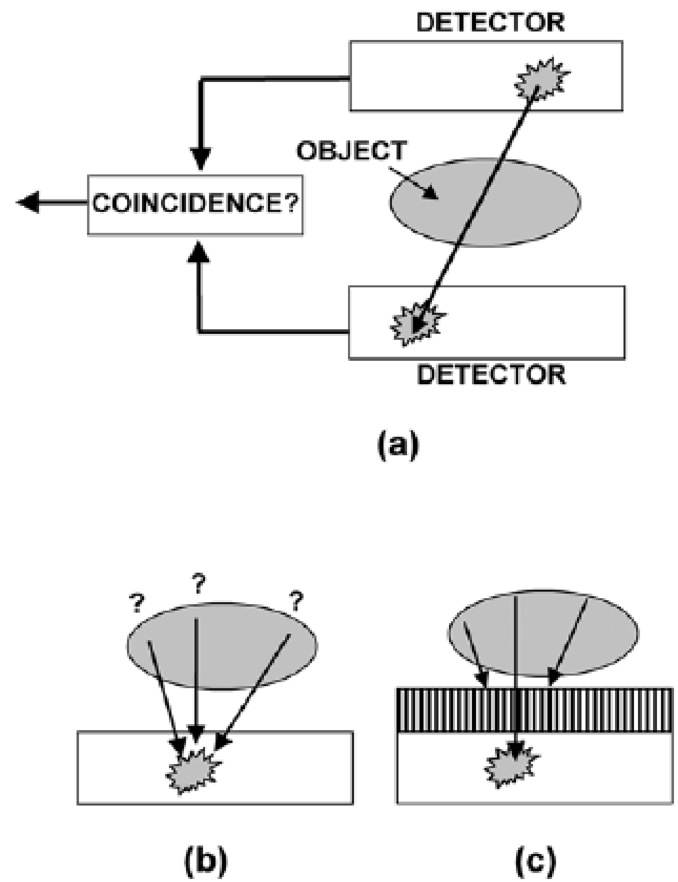


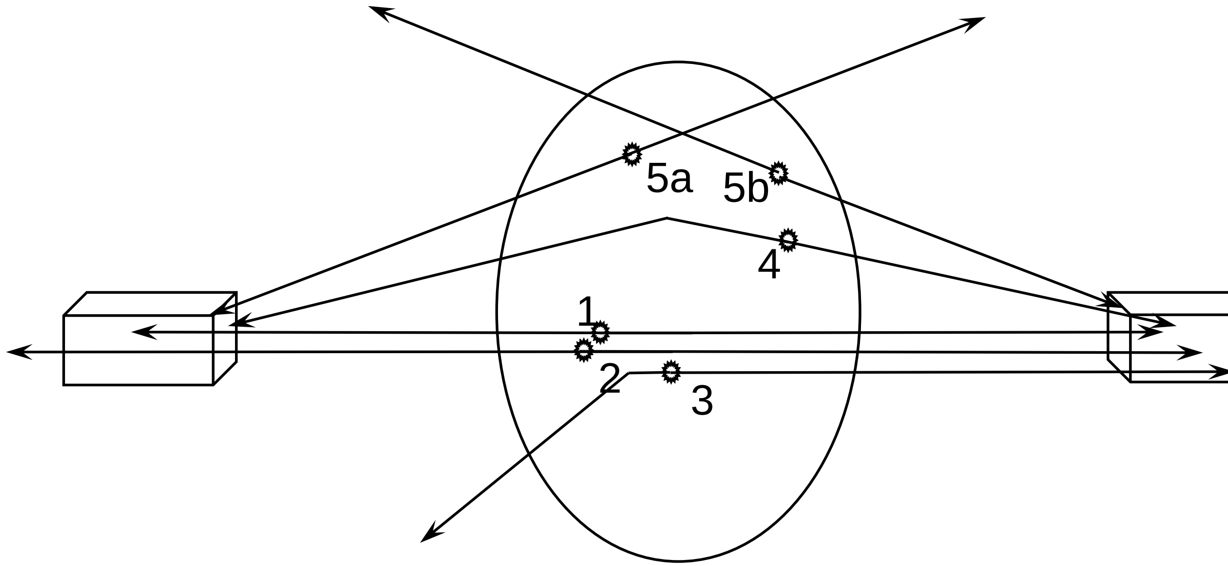

FIGURE 1. The process of positron emission and subsequent positron-electron annihilation results in two 511 keV annihilation photons emitted 180° apart. The site of annihilation is usually very close to the point of positron emission because the emitted positrons rapidly lose their energy in tissue (see Figure 5).

# PET v.s SPECT

**FIGURE 2.** (A) Radionuclides that decay by positron emission result in two annihilation photons emitted 180° apart. If both photons are detected, the detection locations define (to within the distance traveled by the positron prior to annihilation) a line along which the decaying atom was located. (B) Radionuclides that decay by emitting single photons provide no positional information, as a detected event could originate from anywhere in the sample volume. (C) For single photon imaging, physical collimation can be used to absorb all photons except those that are incident on the detector from one particular direction (in this case perpendicular to the detector face), defining a line of origin just like the coincident 511-keV photons do following positron emission. To achieve this localization, however, the radiation from the majority of decays has been absorbed and does not contribute to image formation, leading to the detection of many fewer events for a given amount of radioactivity in the object. Absorptive collimation of this kind is the approach used in planar nuclear medicine imaging and in single photon emission computed tomography (SPECT).



# Coincidence Events




1. Detected True Coincidence Event
2. True Event Lost to Sensitivity or Deadtime
3. True Event Lost to Photon Attenuation
4. Scattered Coincidence Event

5a,b. Random Coincidence Event

# Coincidence Even

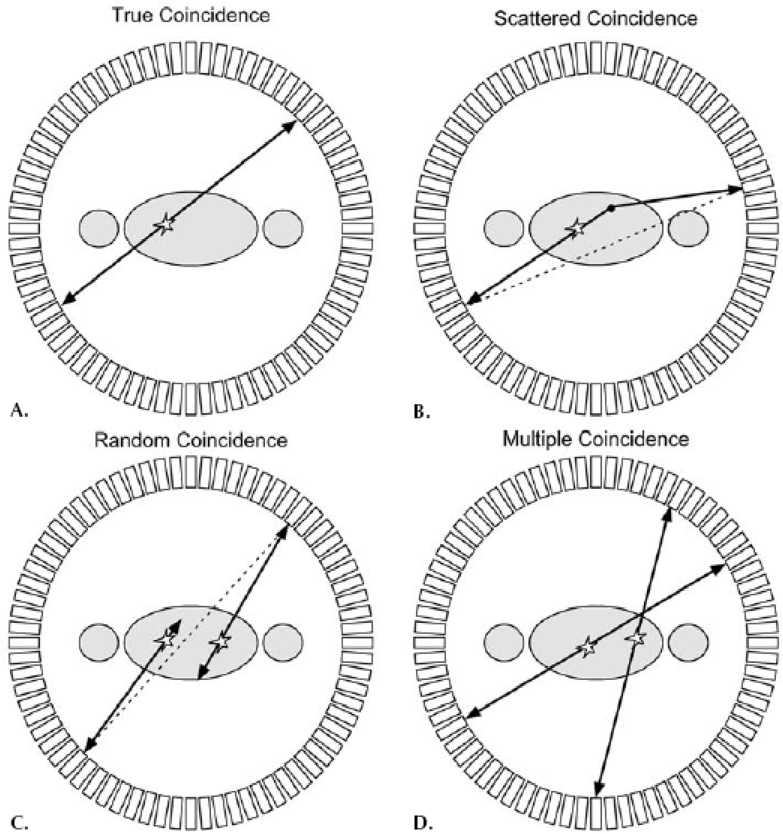
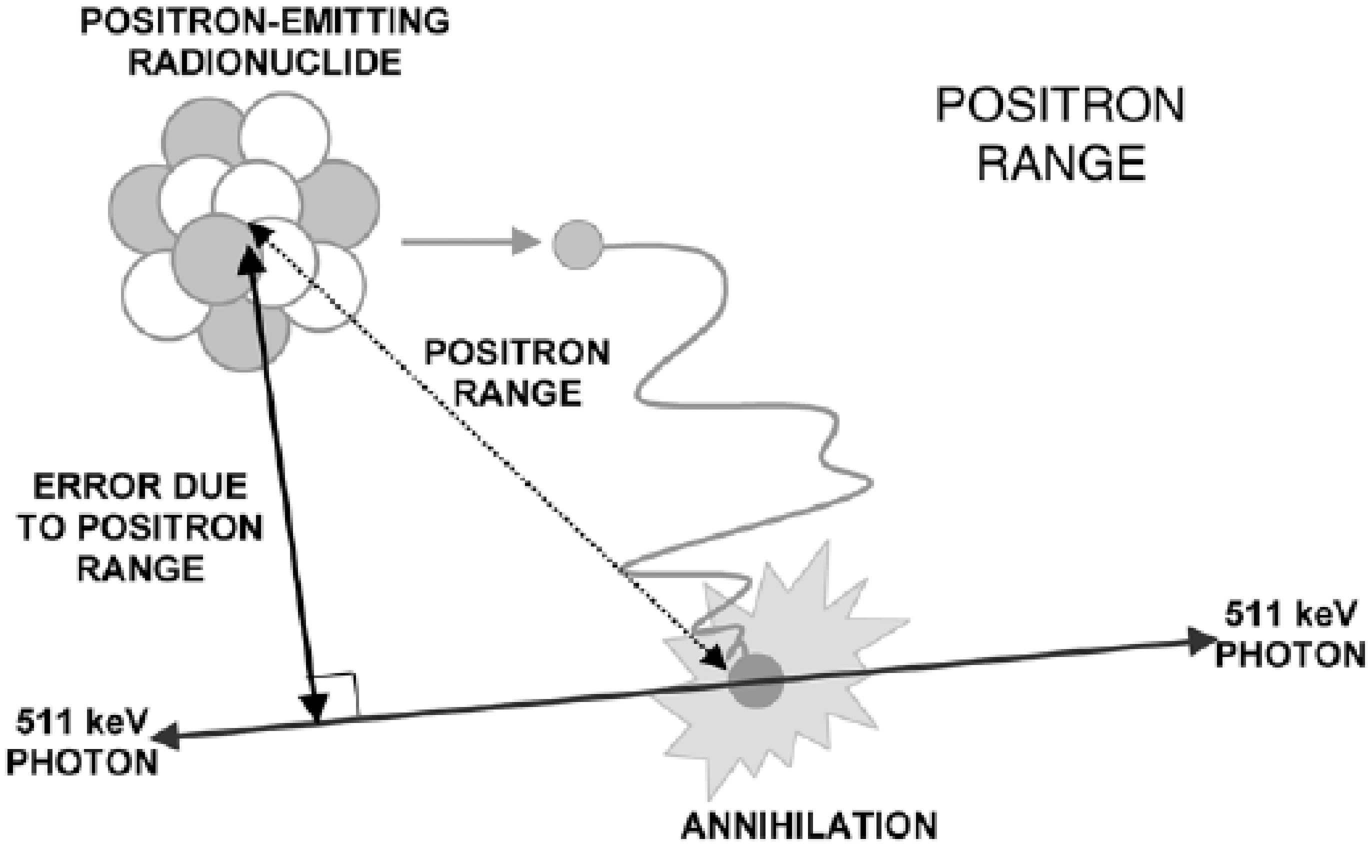



FIGURE 22. Illustration of the four main coincidence event types. A: True coincidence. Both annihilation photons escape the body and are recorded by a pair of detectors. B: Scattered coincidence. One or both of the two annihilation photons interacts in the body prior to detection. This results in a mispositioning of the event. C: Random coincidence: A coincidence is generated by two photons originating from two separate annihilations. These events form a background in the data that needs to be subtracted. D: Multiple coincidence: Three or more photons are detected simultaneously. Due to the ambiguity of where to position the events, these normally are discarded. (Reprinted from *Physics in Nuclear Medicine*, 2nd ed, Cherry SR, Sorenson JA, Phelps ME, W.B. Saunders, New York 1986, with permission from Elsevier.)

# Physical Limits of Resolution

- Positron Range
  - Positron will travel a short distance before annihilation with electron
  - Isotope dependent with resolution loss being about 0.5-2.0 mm
- Angulation
  - Because of momentum of positron-electron pair annihilation photons are not exactly $180^0$ apart
  - Loss of resolution is a function of distance between detectors
  - X = d/2 tan (0.25) so for 100 cm diameter loss is 2.2 mm
- Detector Size
  - FWHM of "geometrical" resolution is half the detector width

# Positron Range



# Non-colinearity

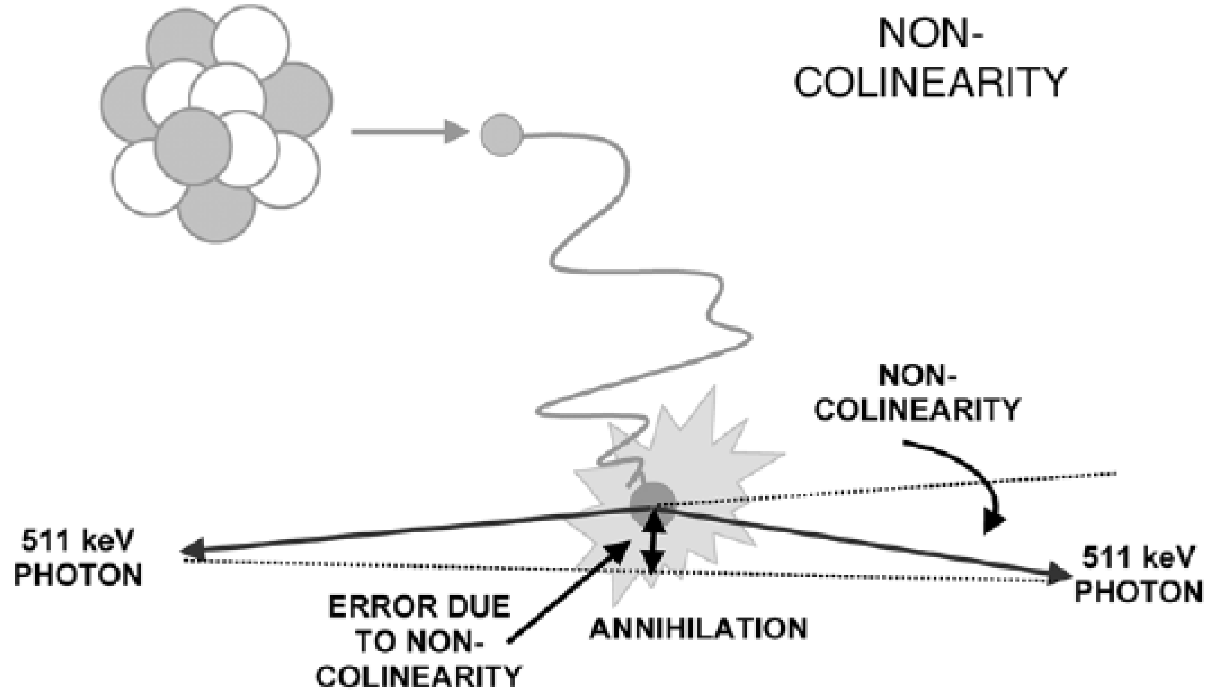



FIGURE 4. Error in determining the location of the emitting nucleus due to positron range (top) and noncolinearity (bottom). The positron range error is dependent on the energy of the emitted positrons. Noncolinearity is independent of radionuclide, and the error is determined by the separation of the detectors. The deviation from noncolinearity is highly exaggerated in the figure; the average angular deviation from 180° is about $\pm$ 0.25°. (Reproduced with permission from Cherry SR, Sorenson JA, Phelps ME. *Physics in Nuclear Medicine*, W.B. Saunders, New York, 2003.)

Assuming a Gaussian distribution and using the fact that the angles are small, the blurring effect due to noncolinearity, $\Delta_{nc}$, can be estimated as:

$$\Delta_{nc} = 0.0022 \times D \quad (10)$$

where $D$ is the diameter of the PET scanner. The error increases linearly as the

# Three types of imaging

- Planar imaging (a single projection similar to X-ray radiograph, 2D)
- Single photon emission CT (SPECT, 3D)
- Positron emission tomography (PET, 3D): a research tool first and now a clinical tool
- Gamma imaging performs a unique role in medical imaging which no other technique can fulfill
- The radiotracer emits $\gamma$-rays with energies in the range 60-511 keV

# Count Rate

- Sensitivity of a gamma camera
- cps/MBq
- 10-50% transmission depending on the depth
- The collimator transmit less than 0.1% of the incident photons and it’s the main factor limiting the overall detection efficiency of the gamma camera.

# 影像的特徵

- 解析度(Resolution)
- 對比(Contrast)
- 雜訊(Noise)
- 靈敏度(Sensitivity, cps/MBq)
- 信雜比(<u>S</u>ignal to <u>N</u>oise <u>R</u>atio, SNR)

# Image Reconstruction From Projections

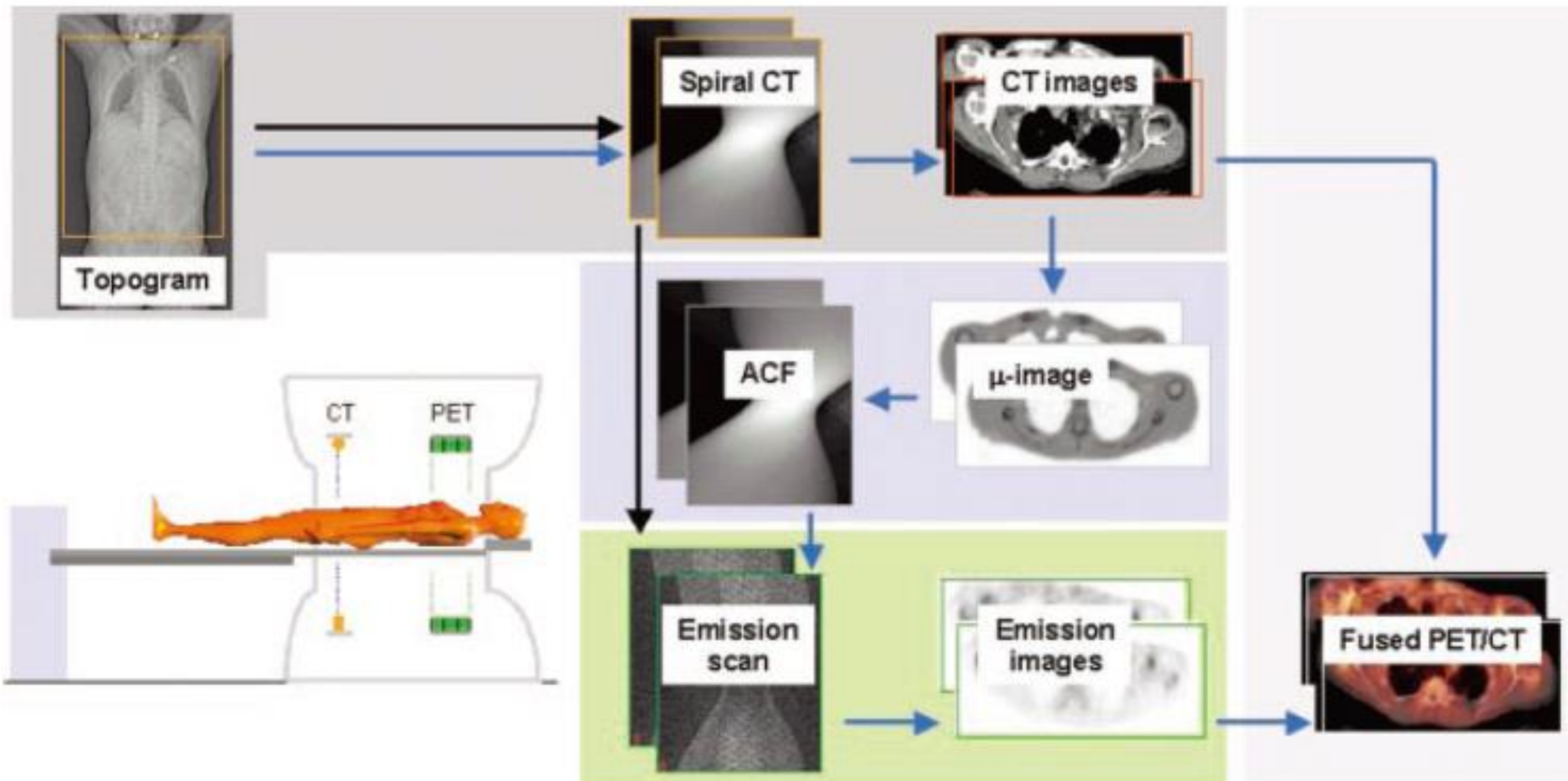
**Three general approaches**

– Simple backprojection

– Analytical techniques (Nobel prize)
*(solution of the inverse Radon transform problem)*
Filtered of the backprojection
Backprojection of the filtered projection
(filtered backprojection: FBP)

– Iterative reconstruction techniques

# Standard PET/CT Imaging Protocol




J Nucl Med 2004; 45:25S-35S

# PET 衰減校正

Blank scan


Measure rod intensity with no object present.

$I_o$

Transmission scan


Measure transmission through the object.

$I_o\, e^{-\int m dl}$

Emission scan


Measure emission from object.

$E_{meas} = E_{true}\, e^{-\int m dl}$

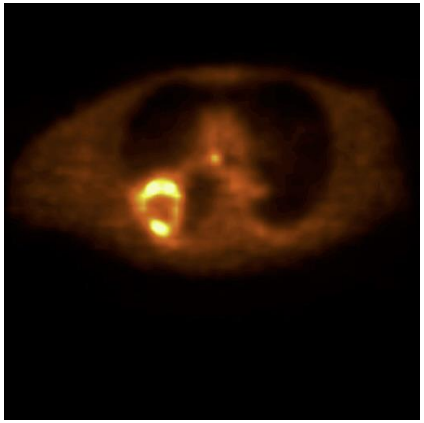
Smooth and take the ratio.

Blank/Transmission = $e^{+\int m dl}$ = ACF

Correct the emission data.

$E_{true} = E_{meas}$ x ACF

# PET/CT



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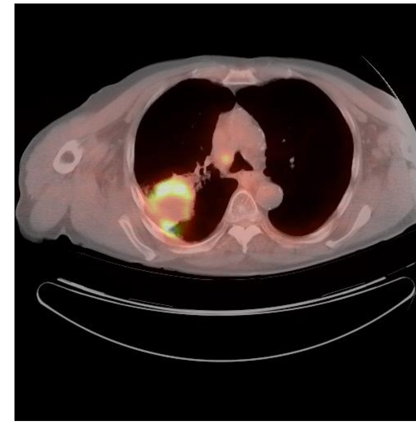
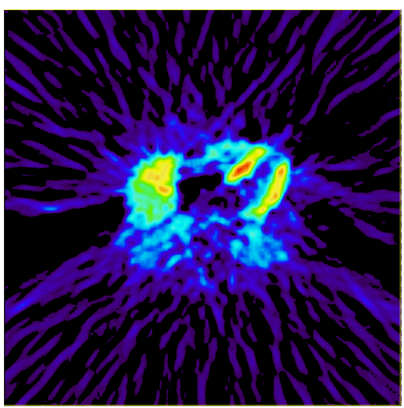


Functional
(PET)

Anatomy
(CT)

Functional and Anatomy



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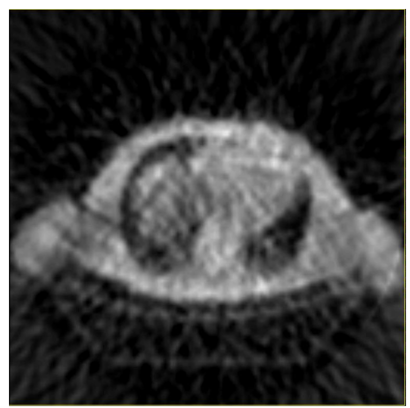

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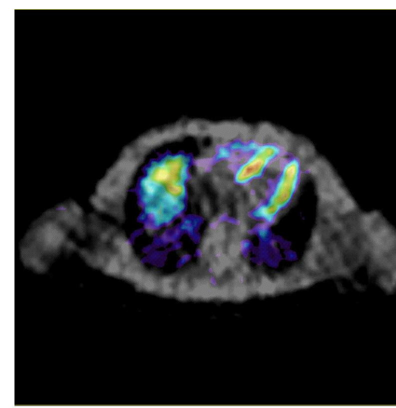

Emission

Transmission

Image Fusion

# PET

# 數位影像的組成

- Matrix (m x n, 256x256)
- Picture element (pixel)
- Image depth (k byte, 1)
- Image storage (m x n x k byte)

  (256 x 256 x 1 byte)

# 核醫電腦資訊

- 8 bit=1 byte, 16 bit= 1 word
- 通常核醫收集影像大多用byte mode或是word mode收集影像資料
- 1 byte=$2^8$ bit =256 (0-255)
- 1 word= $2^{16}$ bit=65536 (0-65535)
- 1 word = 2 byte

| | 優點 | 缺點 | 主要應用 |
|---|---|---|---|
| Byte mode | Less memory, fast | Dead time, truncation error, overflow | Low count studies |
| Word mode | No dead time, no truncation | More memory | High count studies |

# Digital Image

- A digital image consists of a grid or matrix of pixel
- Pixel 為數位影像之最小單位
- 電腦記憶體容量單位

  $1k=2^{10}=10^3$ ； $1\ Mega=2^{20}=10^6$

  $1\ Giga=2^{30}=10^9$ ; $1\ Tera=2^{40}=10^{12}$

# Digital Number

- 通常用二進位或十進位來表示每個pixel的count數，以及even location

Ex. {13} 在十進位=$(1*10^1)+(3*10^0)$

在二進位=$(1*2^3)+(0*2^2)+(1*2^1)+(1*2^0)$，此為轉換為十進位的算法。Binary number就是1011，binary number裡面，每一個digital就叫做1 bit。（所以1011就是4 bit）

Thank you for your attention