PRINCIPE DE FONCTIONNEMENT DU PET-SCAN

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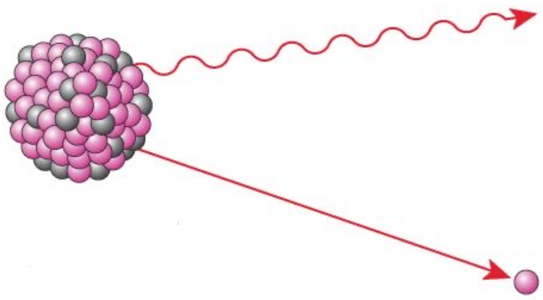
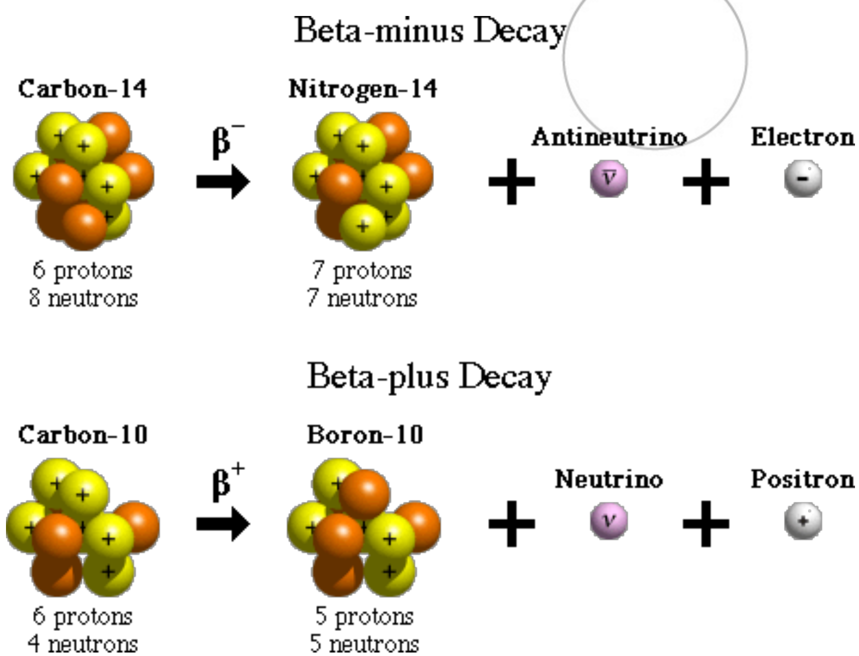
Basics - reminder



β decay



$$^{14}_{6}C \rightarrow {}^{14}_{7}N + \nu_e + e^-$$

$$^{10}_{6}C \rightarrow {}^{10}_{5}B + \nu_e + e^+$$

first physical evidence of the neutrino

continuous energy spectrum for the e^- (keV – MeV)

lower ionizing power and higher range than α

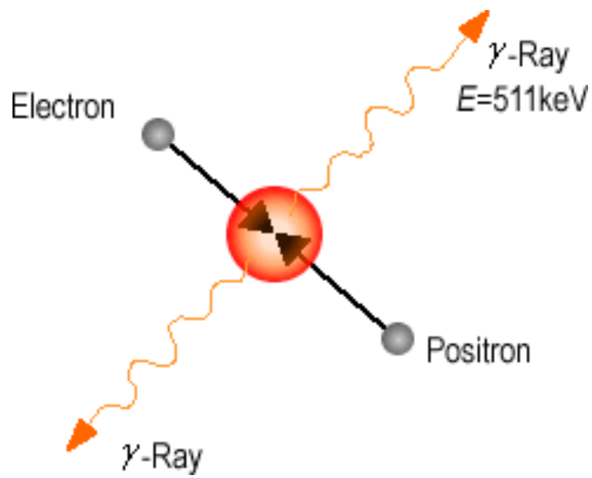
- stopped by a few millimeters of aluminum

β+ emission

emission of an antiparticle : the positron e^+

the positron loses its kinetic energy and quickly annihilates with an atomic electron from the medium

- emission of two 511 keV *(=$m_e c^2$)* gamma photons at ≈ 180°



$$e^+ + e^- \longrightarrow 2\,\gamma$$

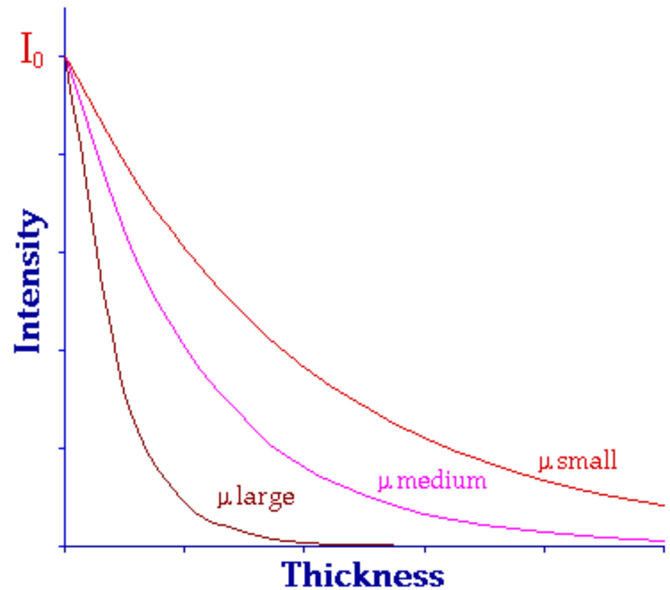
→ allows for coincidence detection
→ fundamentals of PET detection

Interaction of γ rays

attenuation by absorber

- the γ ray intensity decreases exponentially with the length of medium, due to the combined effect of the different interaction modes

$$I(x) = I_0\, e^{-\mu x}$$



- linear attenuation coefficient : probability per unit path length that the γ photon is removed

$$\mu = \tau\ (photoelectric) + \sigma\ (Compton) + \kappa\ (pair)$$

Positron Emission Tomography

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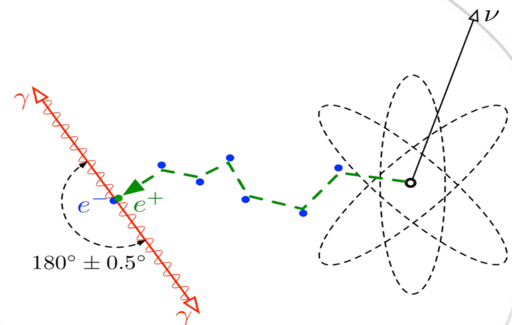
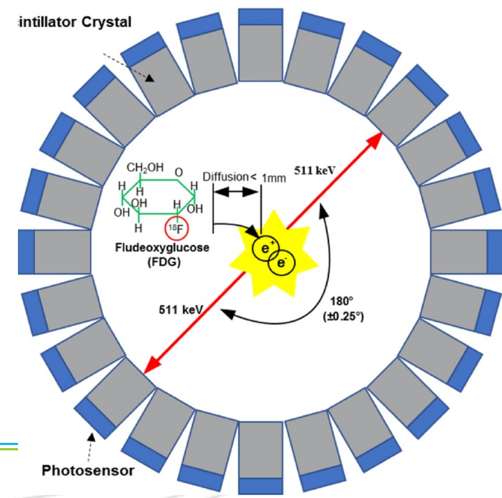
Positron annihilation

Positron emission tomography (PET)

- Technique based on the detection of 511 keV gammas emitted following the annihilation of β+ produced by decay with electrons in the medium

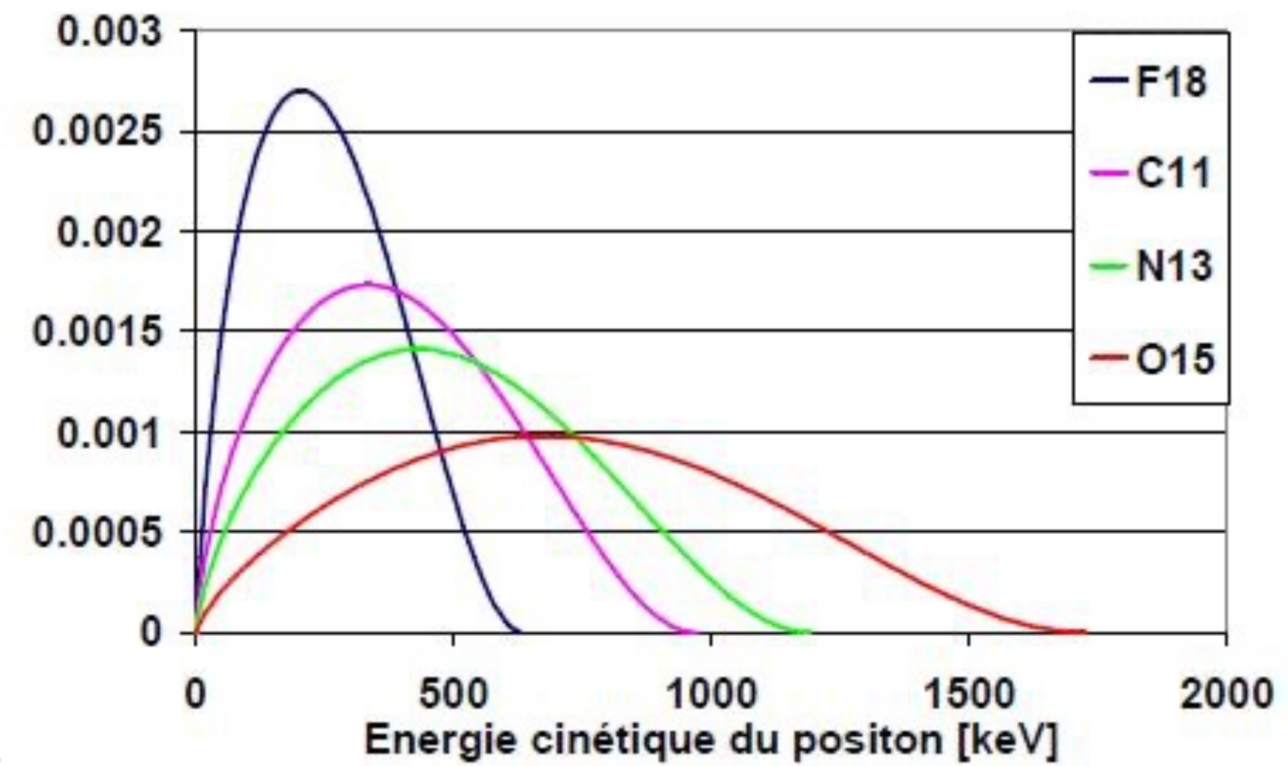
$$ {}^{A}_{Z}X^{*} \longrightarrow {}^{A}_{Z-1}Y + \beta^{+} + \nu $$





Positron annihilation

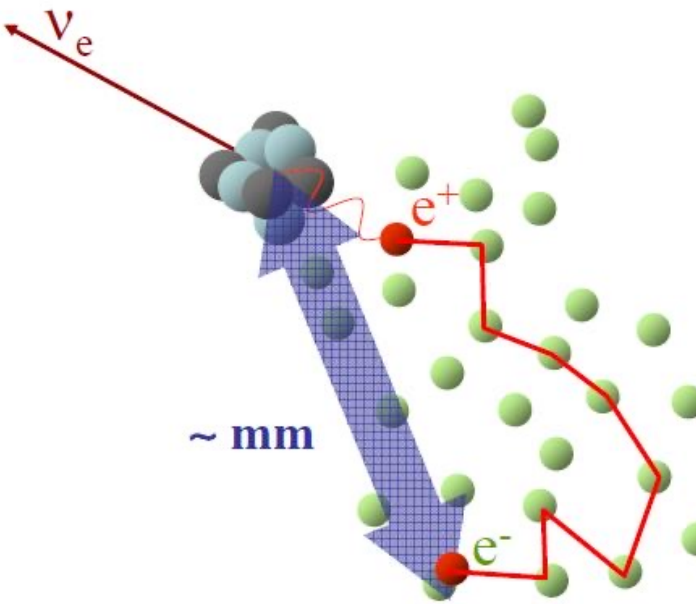
continuous energy spectrum



Positron annihilation

range of the positron

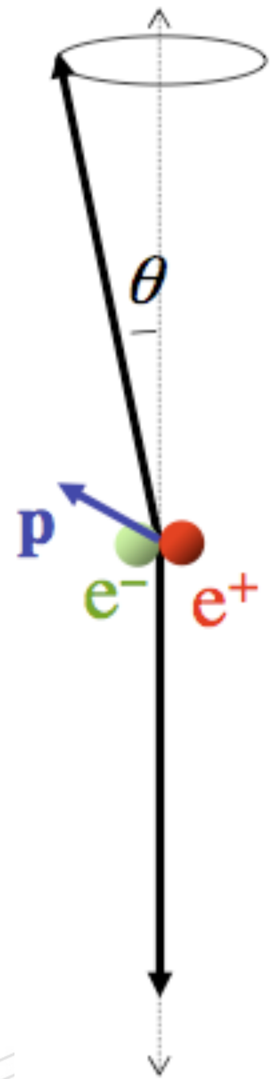
- loses its energy by successive collisions with the material
- distance between the positions of emission and annihilation depends on the energy
- limits the spatial resolution !



Positron annihilation

gamma collinearity

- if p≠0 at the annihilation
  - deviation θ from collinearity
- FWHM = 0.58°



β+ emitters

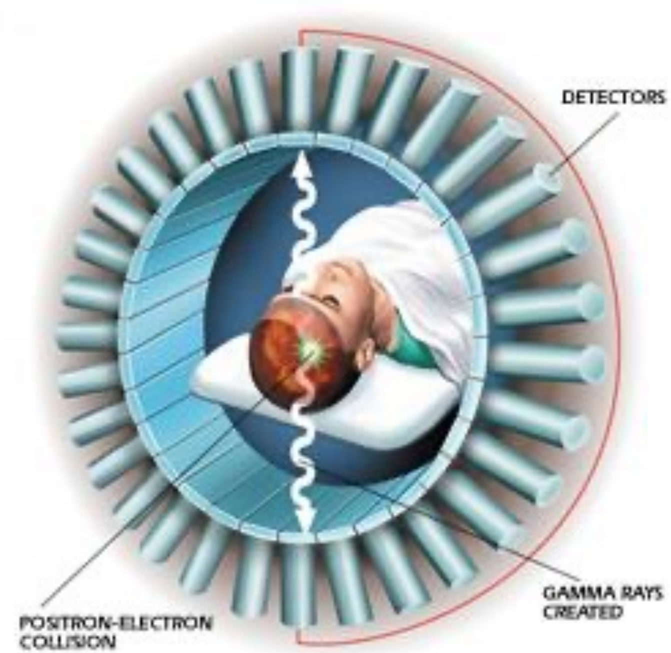
Isotope	Half-life (min)	Positon max energy (MeV)	Path of the positon in water (mm)	
			Mean	Max
^{11}C	20	0.98	1.1	4.1
^{13}N	10	1.19	1.5	5.1
^{15}O	2	1.72	2.7	7.3
^{18}F	110	0.64	0.6	2.4
^{68}Ga	68	1.9	3.1	8.9

Coincidence detection

two events are detected simultaneously

- in the same temporal *coincidence window* (*5-20 ns*)
- allows to determine a line along which the annihilation has occured → same purpose as the collimator in conventional gamma cameras
- often termed *electronic collimation*

→ effect on sensitivity ?



Coincidence detection

the two detectors in which the detection of the photons has taken place provide the line of response (LOR)

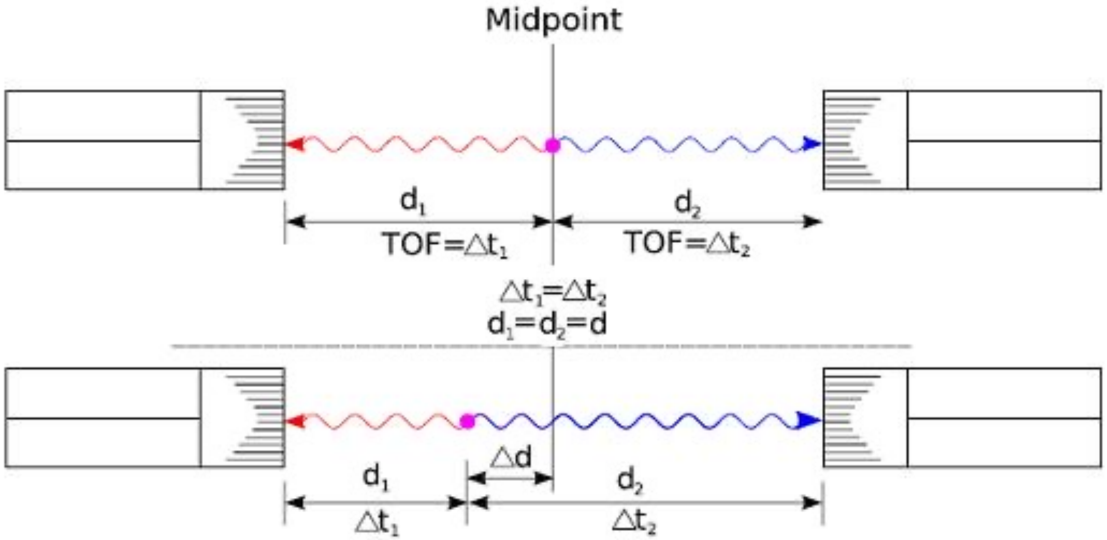
- no direct information on the exact position along that line
- necessity of image *reconstruction* (see Recon chapter)

in theory : with sufficient time resolution, it would be possible to determine the exact point of annihilation of the positron

- *Time-Of-Flight* detection (TOF)

Coincidence detection

Time-Of-Flight



$$d_2 = d + \Delta d$$
$$d_1 = d - \Delta d$$
$$\Rightarrow d_2 - d_1 = 2\Delta d$$
$$\Rightarrow \Delta t_2 - \Delta t_1 = \frac{2\Delta d}{c}$$

c = speed of light = $30\ \frac{cm}{ns}$

$\Rightarrow$ For 500 ps time difference, there is 7.5 cm displacement from the midpoint

Detectors

high stopping power for 511 keV photons

- high Z (photoelectric effect)
- high density

fast decay (high count rates)

- t < 100 ns

affordable

- high number of detectors !

Crystal

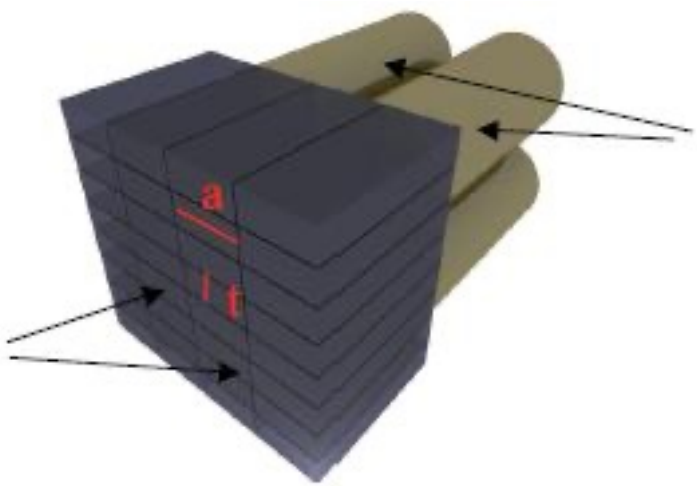
Property	NaI(Tl)	BGO	LSO	GSO	YSO	BaF_2
Density (g/cm3)	3.67	7.13	7.4	6.71	4.53	4.89
Effective *Z*	50.6	74.2	65.5	58.6	34.2	52.2
Decay constant (ns)	230	300	40	60	70	0.6
Light output (photons/keV)	38	6	29	10	46	2
Wavelength (nm)	410	480	420	440	420	220
ΔE/E (%)	6.6	10.2	10	8.5	12.5	11.4
μ/ρ (cm^2/g)	0.3411	0.9496	0.8658	0.6978	0.3875	0.4545
μ (cm^{-1})	0.0948	0.1332	0.117	0.104	0.853	0.0929

- NaI(Tl) : used in the first PET cameras
- BGO : first used in 1979, used in most cameras from 1980 to 2000
- GSO : Philips Allegro system
- LSO : used in modern Siemens cameras
- LYSO : substitute to LSO in Philips TOF cameras

Block detectors

high number of detector elements (crystals) are measured by a smaller number of PM tubes

- e.g. 6 x 6 crystals coupled to 4 PMTs, positioning circuit similar to gamma cameras
- cost reduction
- good spatial resolution and axial coverage

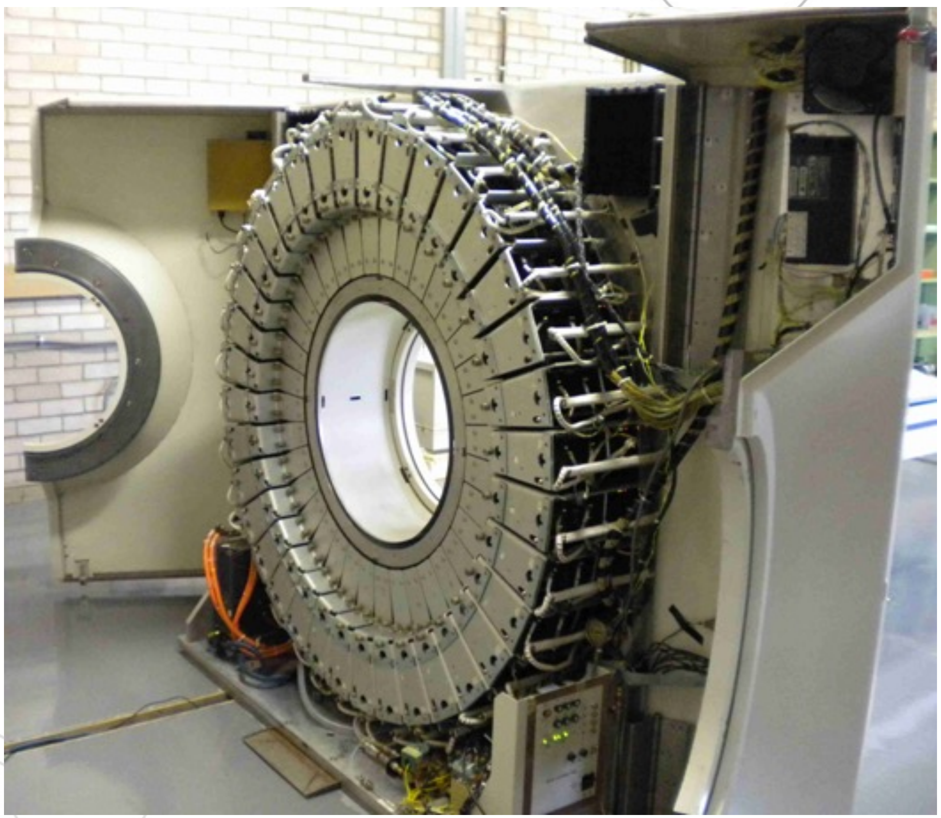


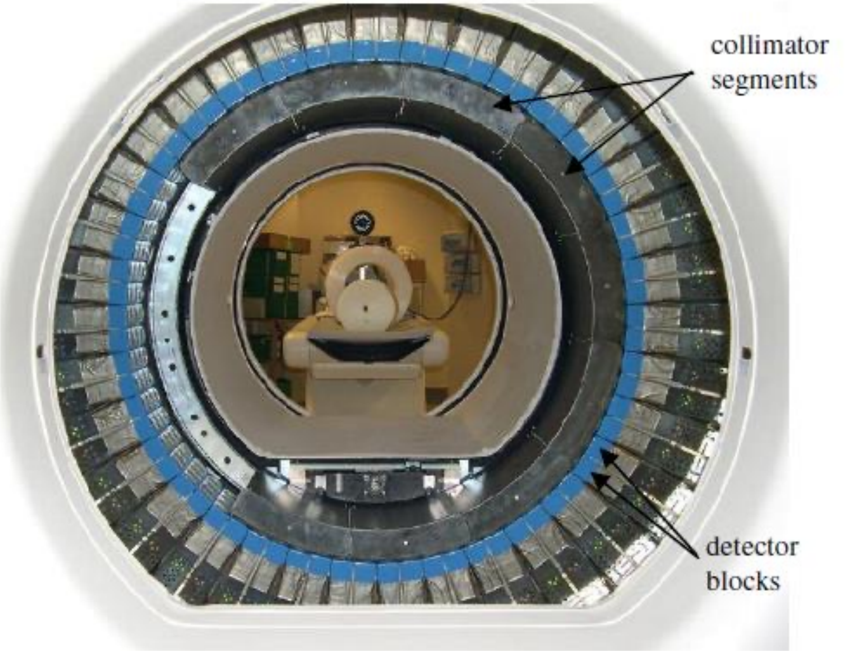
a = 8.4 mm

t = 4.0 mm

e = 30 mm

Scanner design





Data acquisition

event detection relies on electronic collimation

an event is ***valid*** if:

- two photons are detected within the coincidence window
- the LOR between them is within a valid acceptance angle
- the energy deposited in the crystal by both photons is within the selected window (e.g. 425-650 keV)

→ such coincidence events are referred to as **prompts**

→ include various types of events :

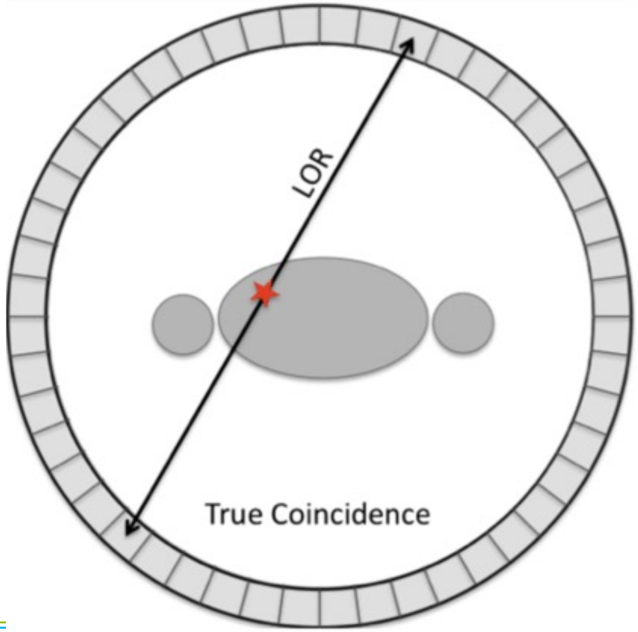
- true
- random
- multiple
- scattered

Data acquisition

Registered events

- True coincidences = Coincidence of two singles coming from the same annihilation and not having scattered in the attenuating medium.

⇒ Well localised on the projection line
⇒ Useful information

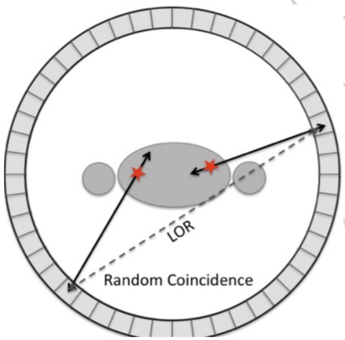


Data acquisition

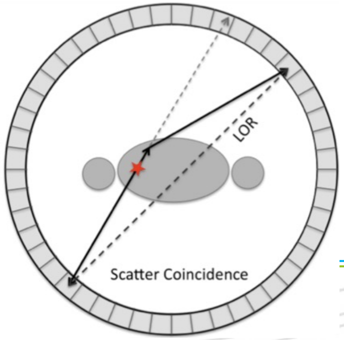
Registered events

- The random coincidences:
  - Two singles not coming from the same annihilation and having diffused or not in the attenuating medium

⇒ Wrong localisation
⇒ Reduction quantitative capacities



- Scattered coincidences:
  - Two singles from the same annihilation, one or both of which have undergone one or more scatterings in the attenuating medium

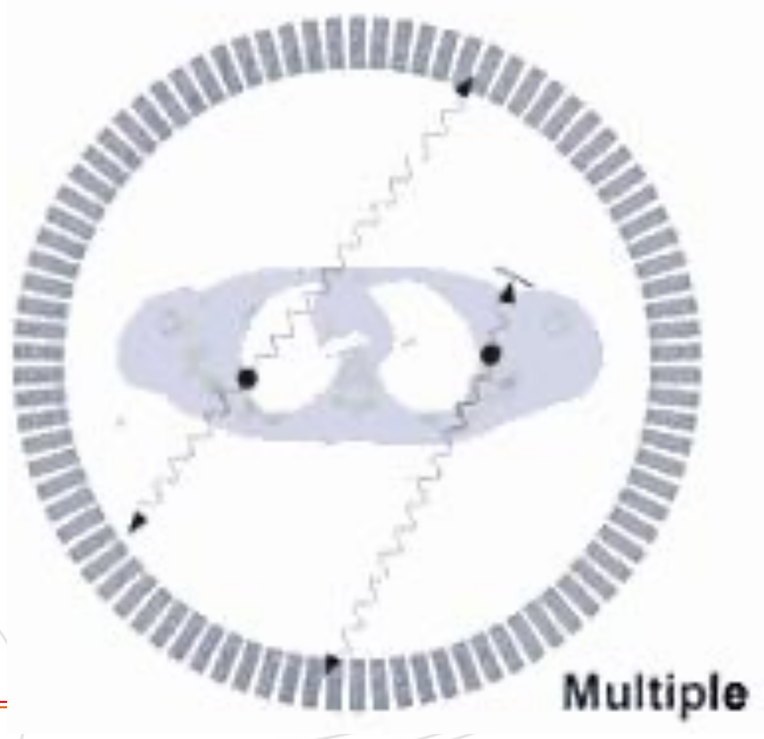


⇒ Wrong localisation
⇒ Lower contrast
⇒ Reduction quantitative capacities

Data acquisition

Registered events

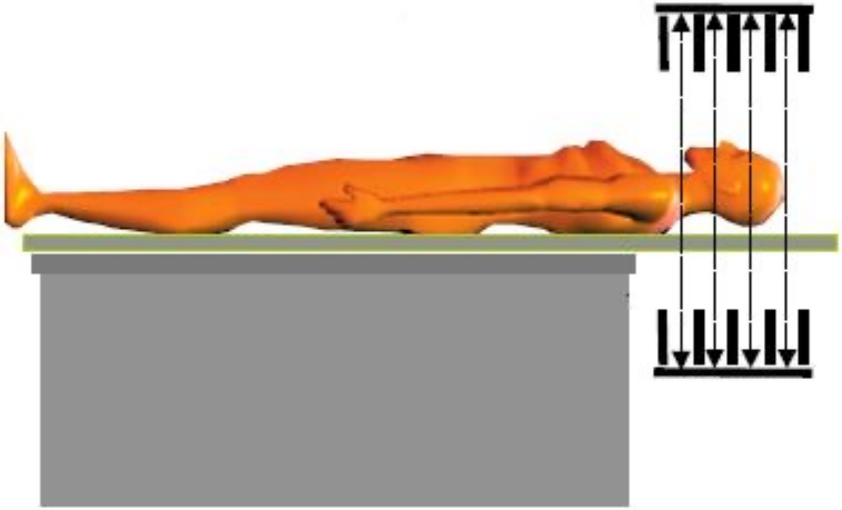
- The multiple coincidences:
  - multiple events are detected → rejected



Acquisition modes - 2D

tungsten septa are placed between the detector rings

- first PET scans
- rejection of oblique events



→ effect on

- spatial resolution and sensitivity ?
- true, random, scatter count rates ?
- reconstruction ?

Acquisition modes - 2D

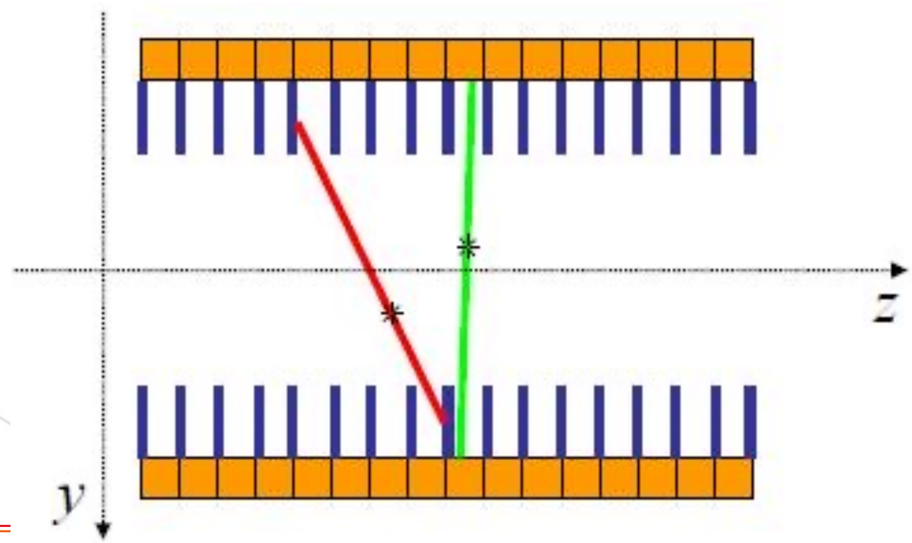
reduction of the scatter and random count rates

better spatial resolution ?

lower sensitivity

easier reconstruction

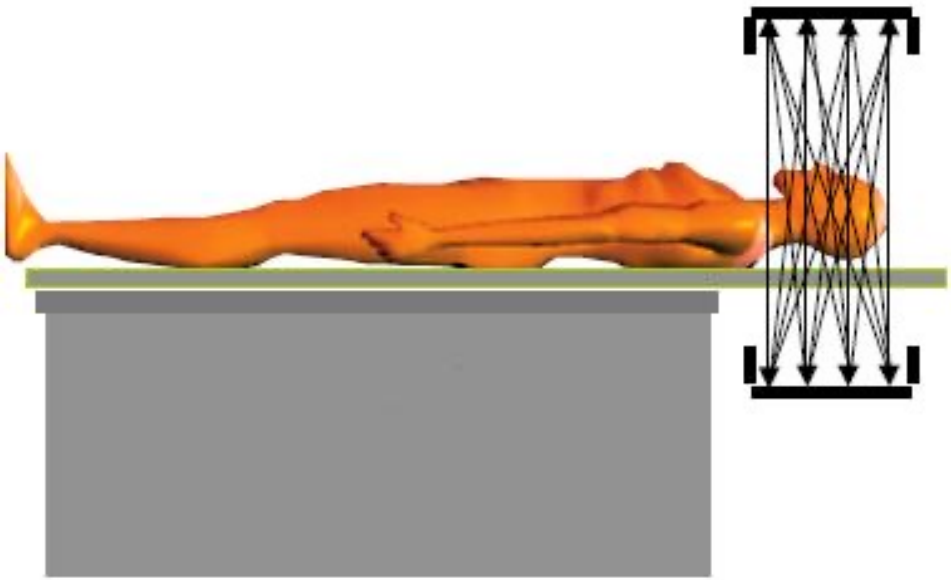
- originally no 3D reconstruction was available



Acquisition modes - 3D

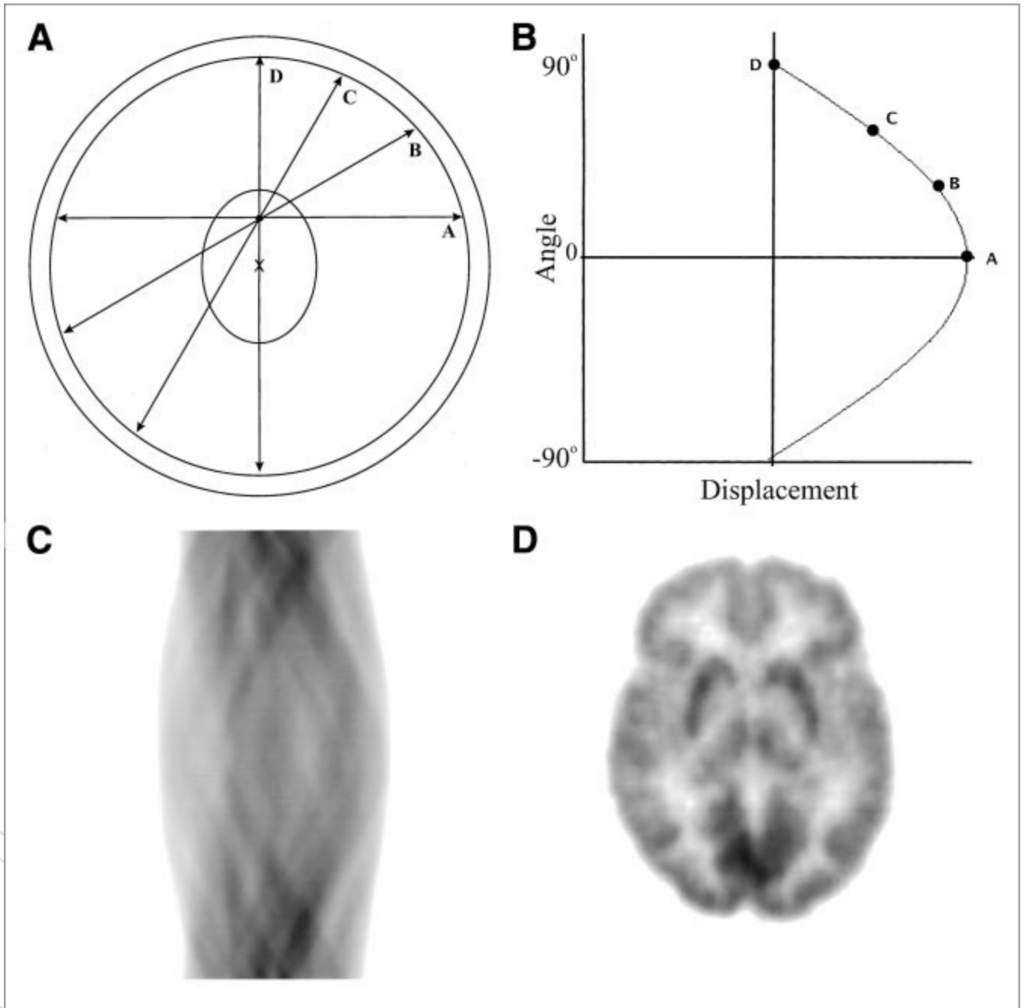
septa are retracted (or absent in modern cameras)

- more LORs are accepted
- if all LORs are accepted : Nc −1 slices



→ scatter correction needed !

Sinogram



resentation of raw coincidence data

e point in the sinogram for each possible LOR

e pixel value is incremented every time a
ncidence event is detected

e sinogram for each position of the axial
npling (each slice)

Sinogram

PET raw data are acquired into sinograms

the LOR associated with each coincidence detection is plotted as a function of angle of orientation versus the shortest distance between the LOR and the center of the gantry

a separate sinogram for each slice

each pixel in the sinogram corresponds to a particular LOR

events associated with a particular detector are plotted on diagonals across the sinogram

Common procedure

F-18 FDG scan

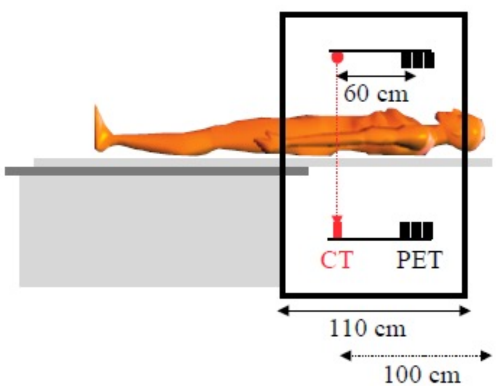
- represents the majority of PET exams (90 %)
- FDG is a glucose analogue that is highly taken up by tumors
- physiologic uptake in brain, heart, liver, muscles, bladder, etc.

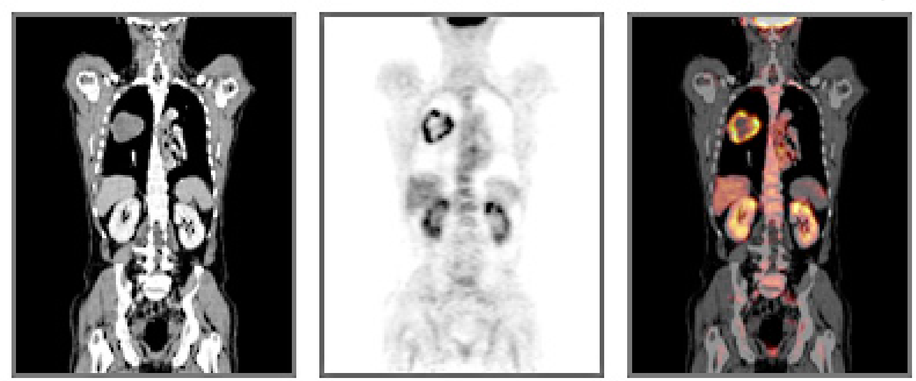
- the patient needs to fast before the exam, and to rest between the injection and start of acquisition to avoid muscle uptake that would degrade the contrast
- the injection-acquisition delay must be reproduced accurately (~ 1 h)
- used for initial staging, follow-up, and therapeutic purposes (e.g. radiotherapy)

Summary

Positron Emission Tomography

- mostly used in oncology (less in neurology and cardiology)
- electronic collimation allows better spatial resolution and quantification than in SPECT techniques (see Quantif chapter)
- modes of acquisition : 2D or 3D, but the 2D tends to disappear
- hybrid PET-CT system : all new devices
- much more expensive than SPECT (number of detectors)





Merci

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