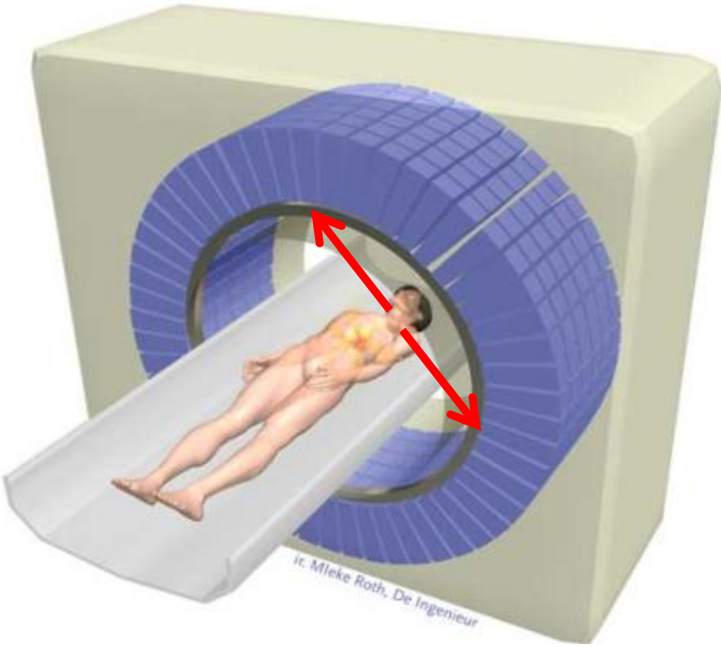
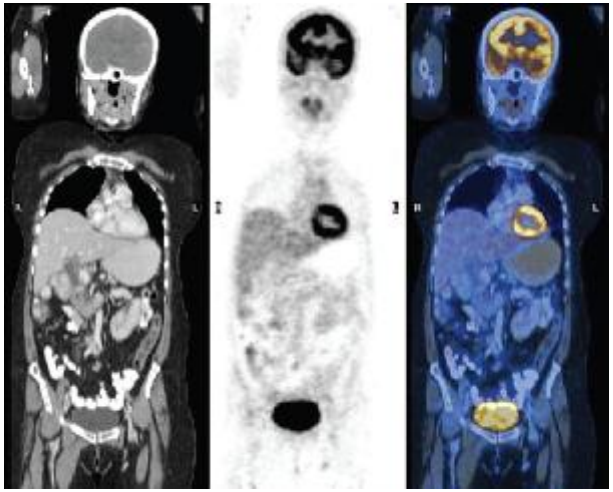




PhD. **Silvano Gnesin**
Institute of Radiation physics (IRA)
UNIL - CHUV

Physics of Positron Emission Tomography (PET)

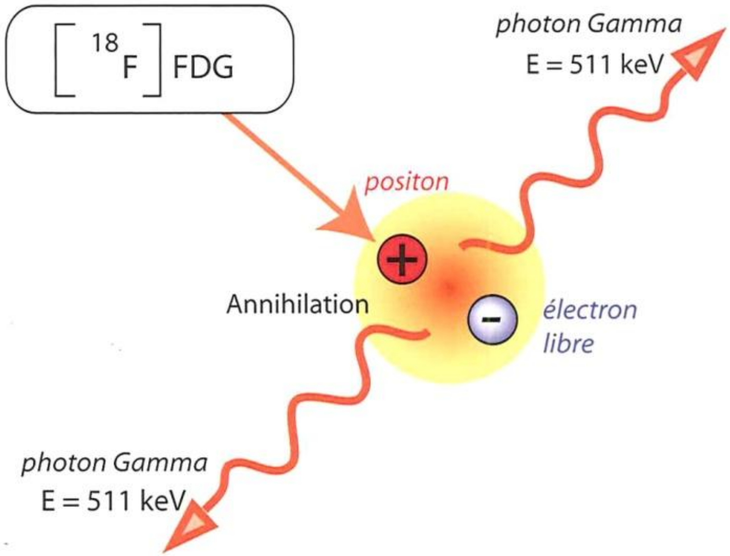
Master of Science EPF-ETH
Degree in Nuclear Engineering and Medical Radiation Physics

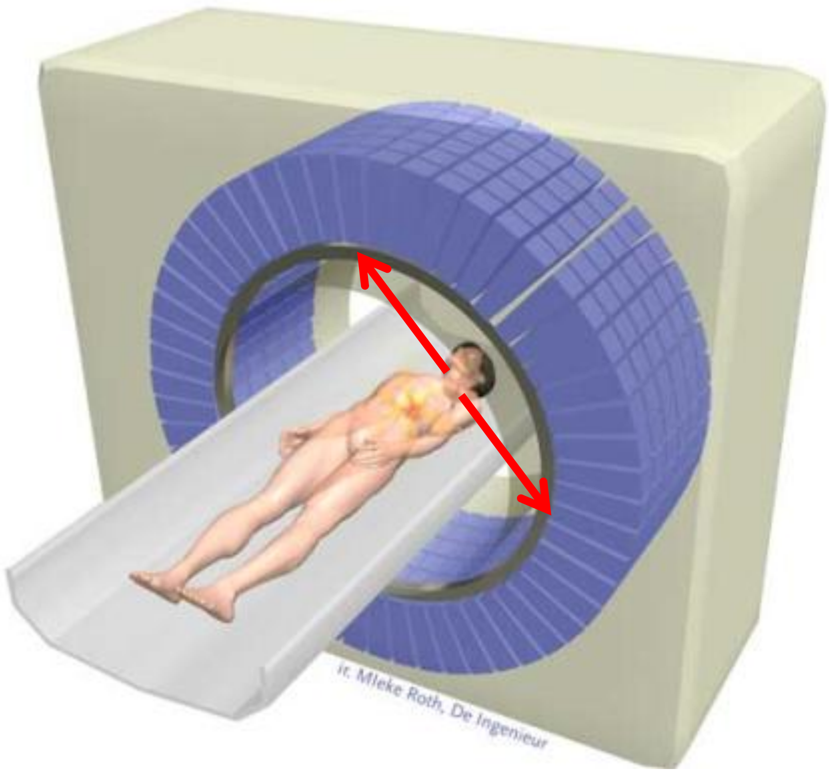
EPFL 30.11.2018

Goals

- Understand and describe **physical bases**, image acquisition and reconstruction processes in **PET**
- Explain **quantitative bias** in **PET** and describe the main **correction methods** :Attenuation, scattering, random coincidence
- Time of flight (TOF) & resolution recovery via point spread function deconvolution
- Mention the parameters influencing image quality (noise contrast, resolution)

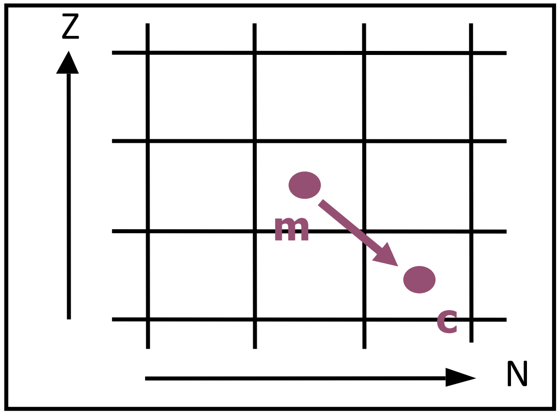
Basic principle of PET





Radionuclides used in PET : ^{11}C, ^{13}N, ^{15}O, **^{18}F**, ^{68}Ga, ^{82}Rb, ^{44}Sc

β^+ Emission

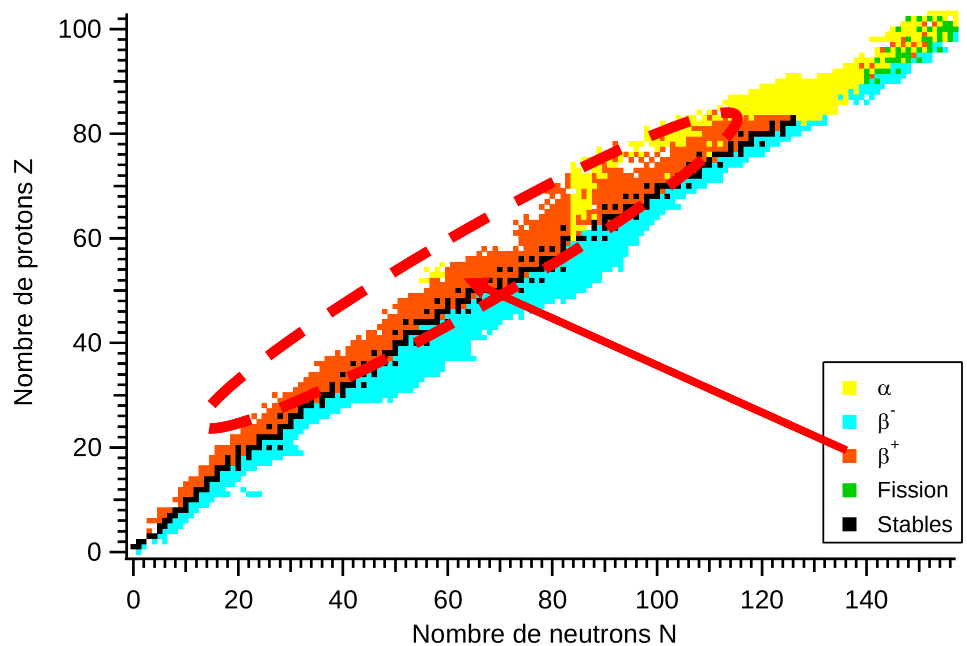


$$^{A}_{Z}E_{N} \rightarrow {}^{A}_{Z-1}E_{N+1} + \beta^{+} + \upsilon$$

positron neutrino

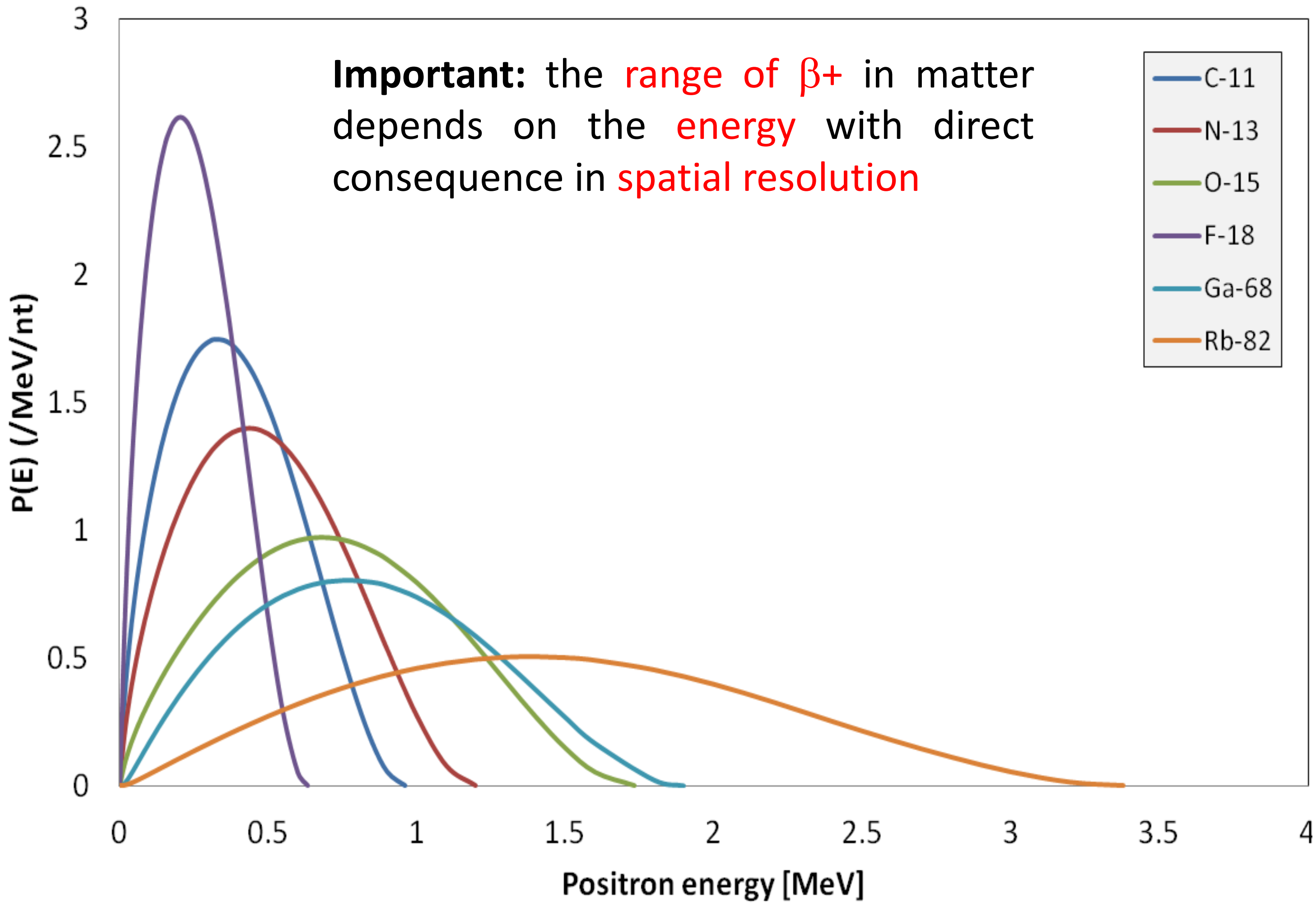
Proton decay

^{11}C, ^{13}N, ^{15}O, ^{18}F, ^{68}Ga, ^{82}Rb, ^{124}I



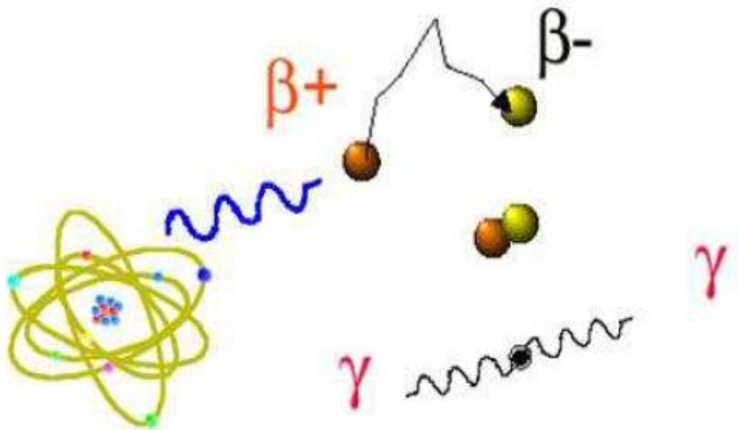

Cyclotron production : ^{18}O (p,n) ^{18}F

Energy spectra of β^+ decay

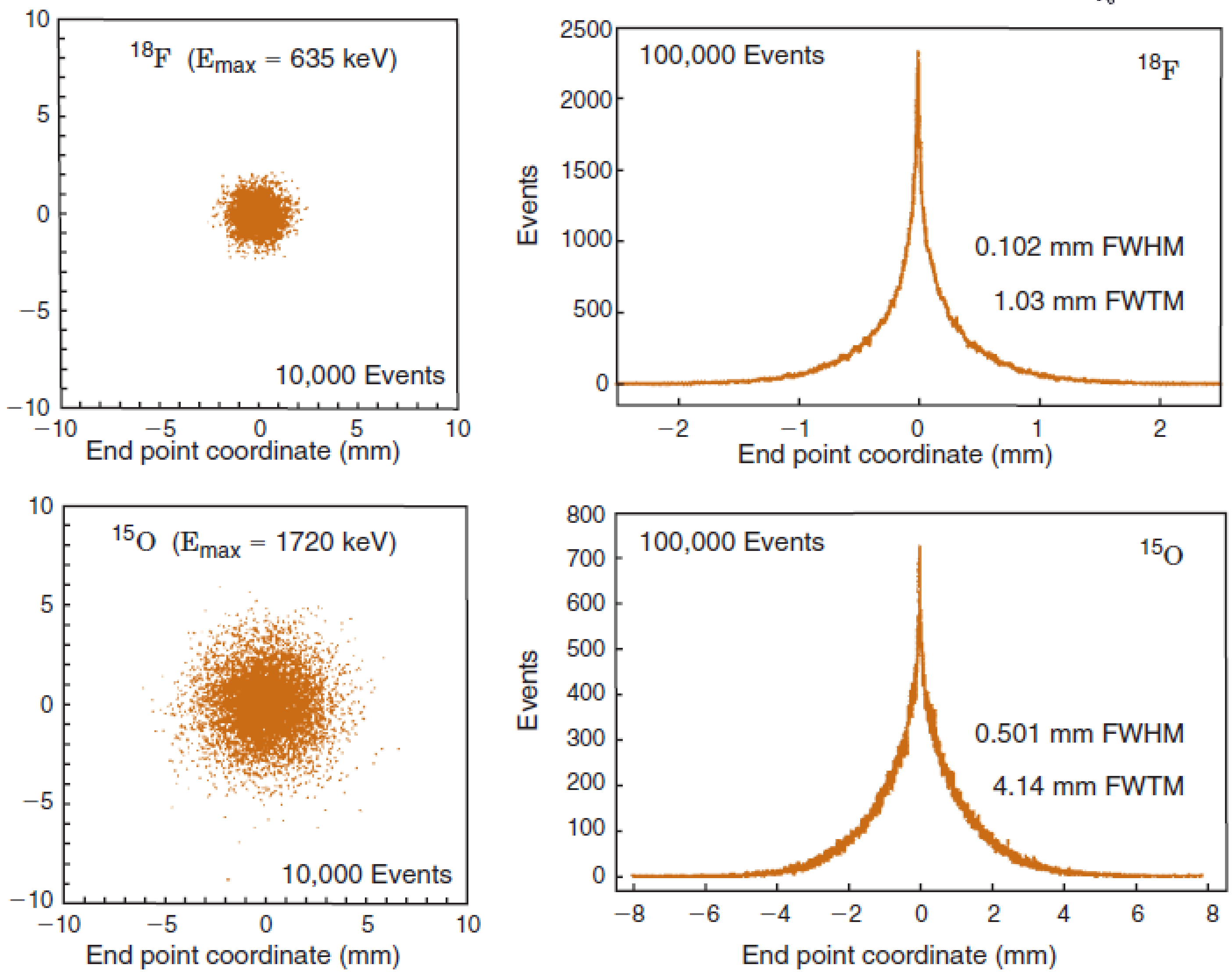


Physical properties of β^+ decay

Radionuclide range	Half-life	$E^+_{\beta,max}$ (MeV)	Max. β^+ range (mm) in water	Average β^+ (mm) in water
^{11}C	20.4 min	0.97	3.8	0.85
^{13}N	10 min	1.20	5.0	1.15
^{15}O	2 min	1.74	8.0	1.80
^{18}F	110 min	0.64	2.2	0.46
^{68}Ga	68 min	1.90	9.0	2.15
^{82}Rb	75 sec	3.35	15.5	4.10

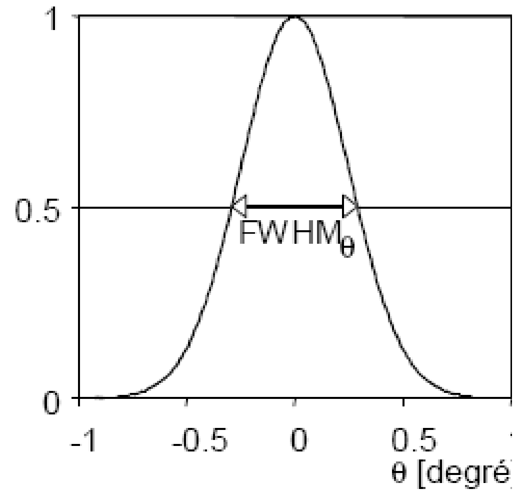


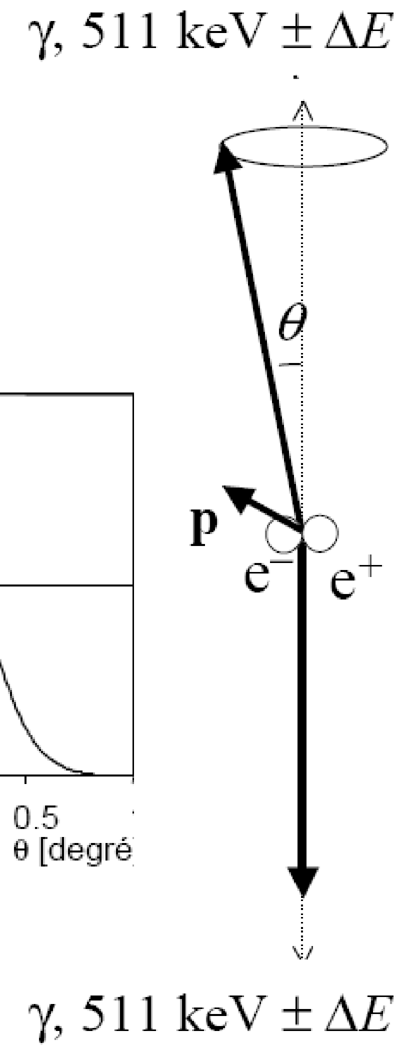
β+ emission: Energy and range



Annihilation gamma emission and directions

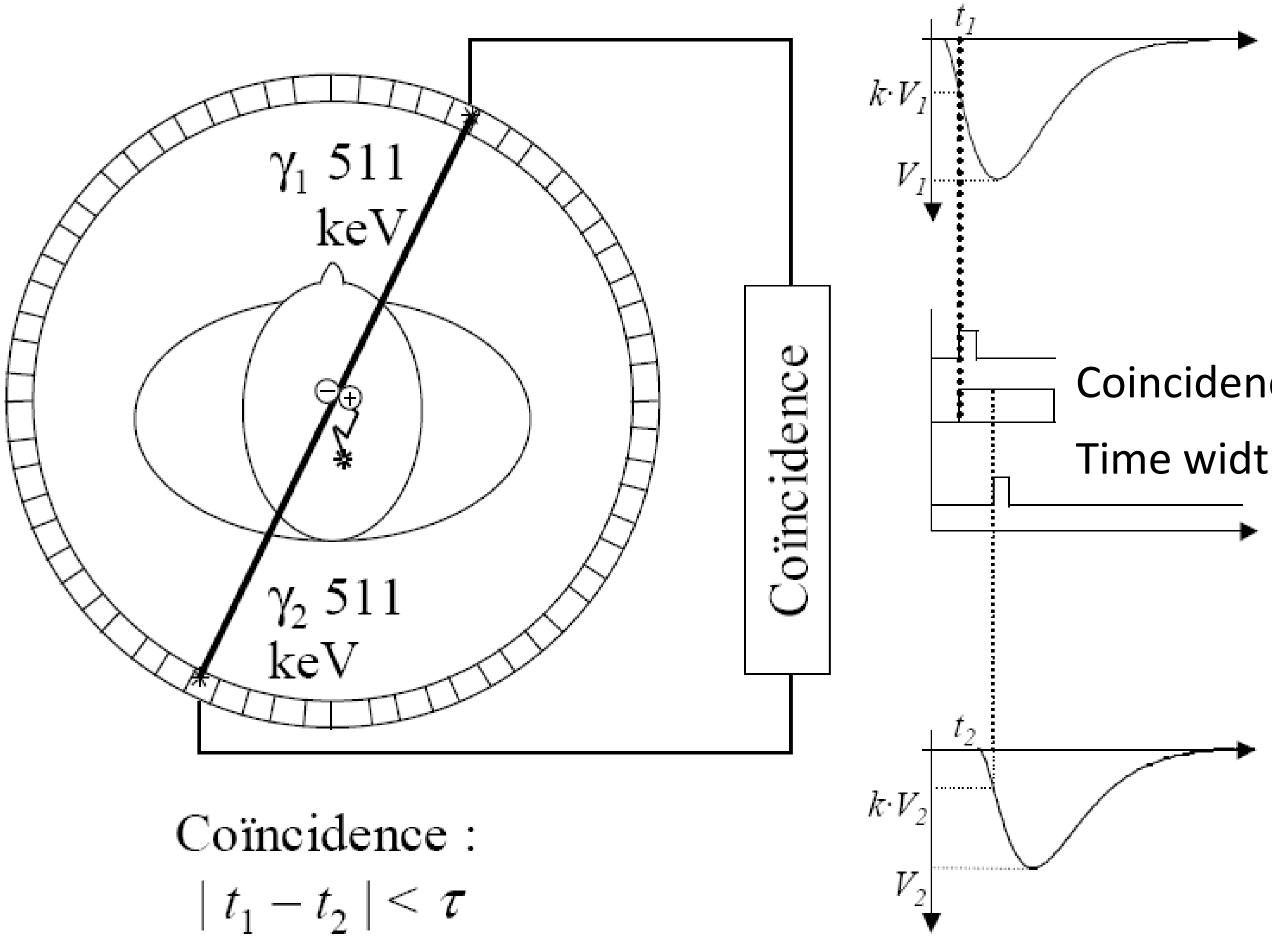
- Positron at rest: annihilates with an atomic electron of the surrounding medium
- Energy release: $2 \times m_e c^2 = 2 \times 511$ keV
- Emission of 2 gamma rays (511 keV) in opposite directions (if total momentum e^+e^- is $p = 0$)
- **Residual momentum**
  - **p** $\neq 0$
  - Non perfect co-linearity $\theta \neq 180$
  - $FWHM_\theta = 0.58°$





Coincidence detection: Electronic collimation

Camera: Detector ring with detector blocs

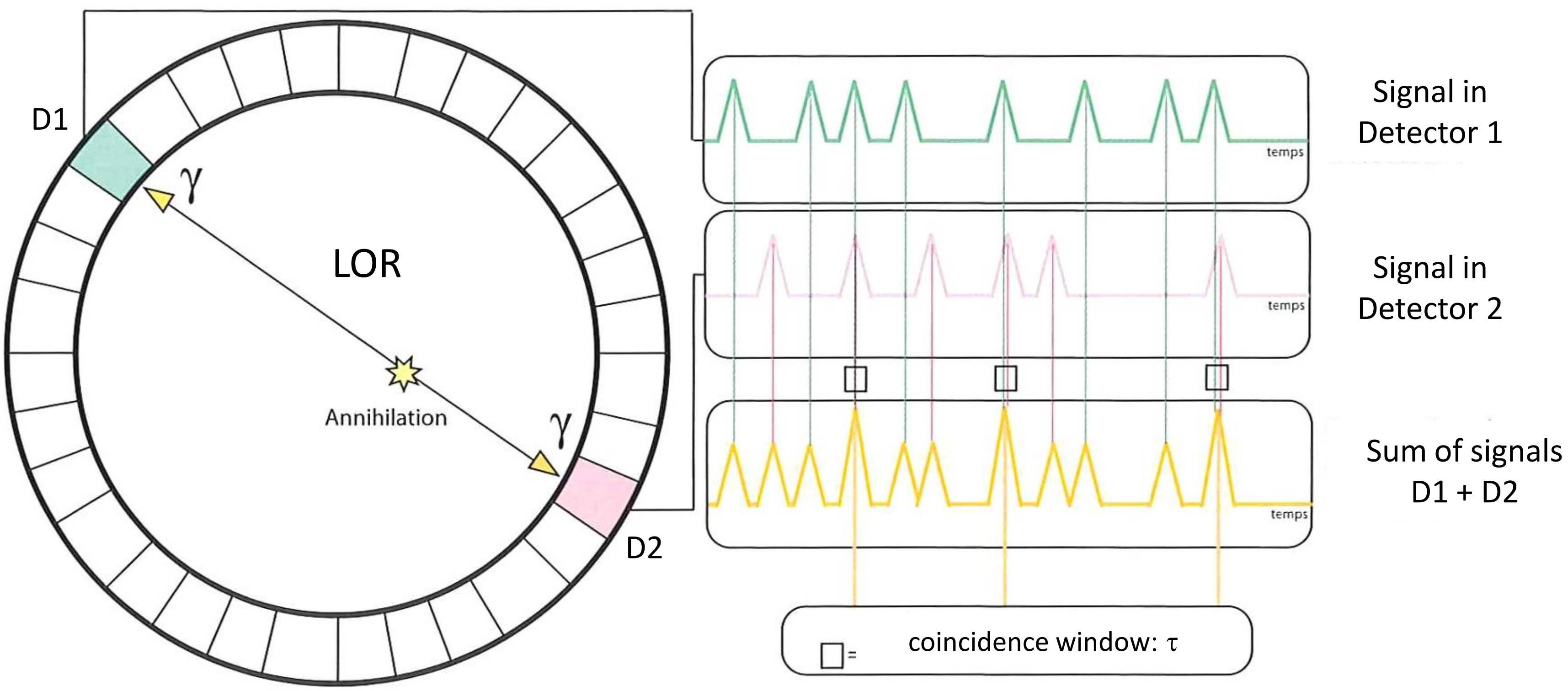


Coincidence time window

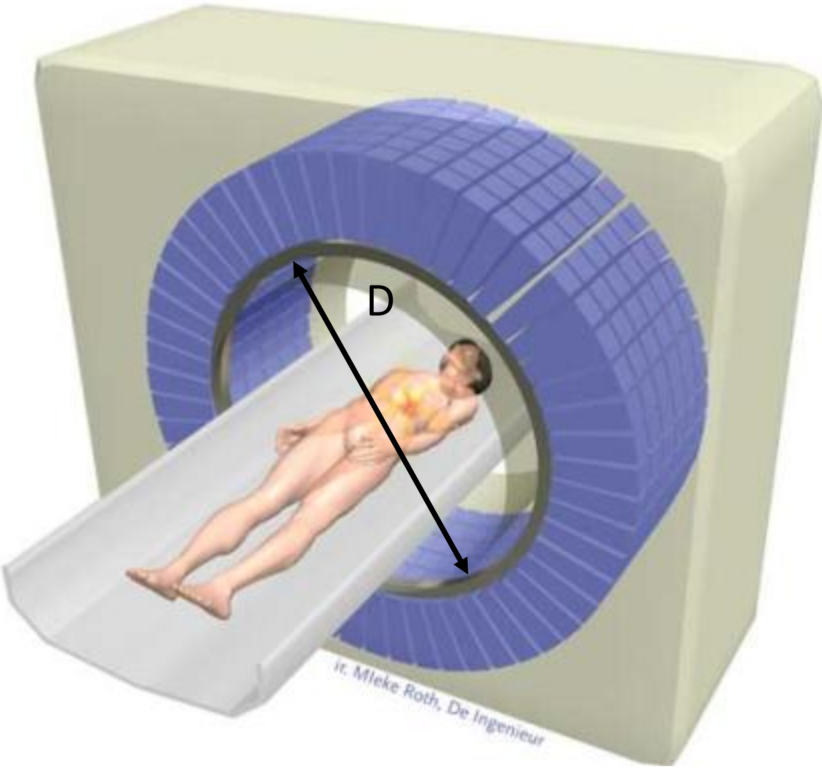
Time width: τ ~ 5-10 ns

Coincidence detection

Only photons detected in coincidence (time and energy) are considered in each LOR



Definition of the time coincidence width



Maximum photon path along
a line of response: D

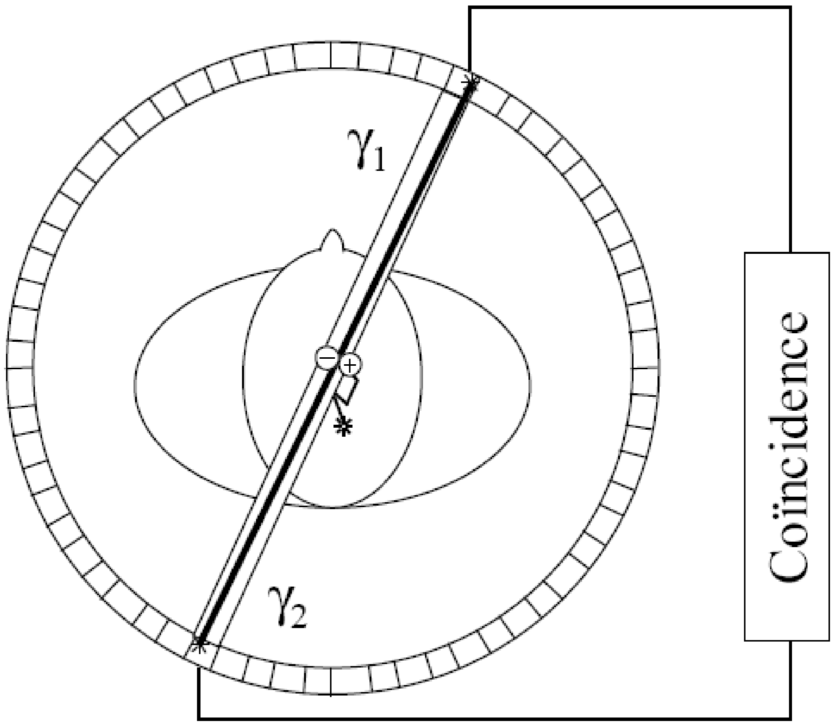
Time need for a photon to travel a distance D;
T = D/c

Practical example: D = 0.8m
→ T = 0.8 m / 3E+8m/s = 2.7E-9s
→ **T ~ 3ns**

Taking into considerations for scintillation light propagation and electric signal collection
→ **T~5ns**

Coincidence detection

LOR : Line Of Response



- **Electronic collimation**
  - Absence of physical collimation (no collimator)
  - The LOR is defined by the tube connecting the two detectors elements responding in coincidence.
- **Acquisition**
  - The number of coincidence recorded for each couple of detectors is scored by the digital acquisition.

We only know the annihilation event happens somewhere along the LOR

PET detectors: main parameters

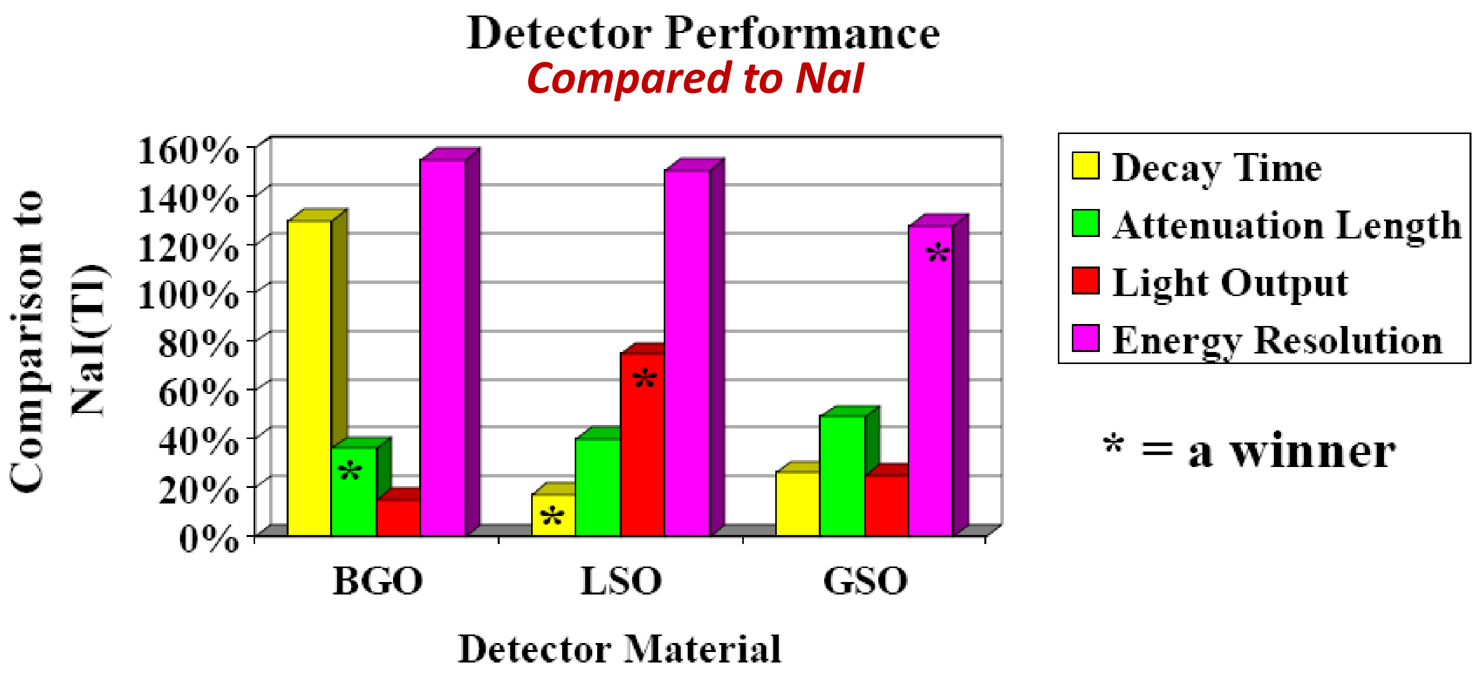
- High detection efficiency at 511 keV
  - Coincidence detection efficiency proportional to $(\text{efficacité}_{\text{détecteurs}})^2$
    - High Z (maximize photoelectric effect)
    - High density materials
- Fast detection
  - Minimize coincidence dead time ($\sim$ 2 x integration time)
    - Detector dead-time < 100ns

PET detectors: main Physical parameters

	NaI (T1)	*BGO*	*BaF$_2$*	*GSO*	*LSO* *	*LuAp*
Densité (g/cm^3)	3,67	7,13	4,9	6,71	7,35	8,34
Z_{eff}	50	73		58	65	65
μ à 511 keV	0,38	0,90	0,45	0,67	0,80	0,91
Photofraction (%)	18	42	19	26	33	32
Constante de temps (ns)	230	300	0,8 630	60 (90 %) 600 (10 %)	40	18
Rendement lumineux relatif	100	22	5 21	20	75	25

*** LYSO performance are similar to LSO**

PET detectors: main Physical parameters



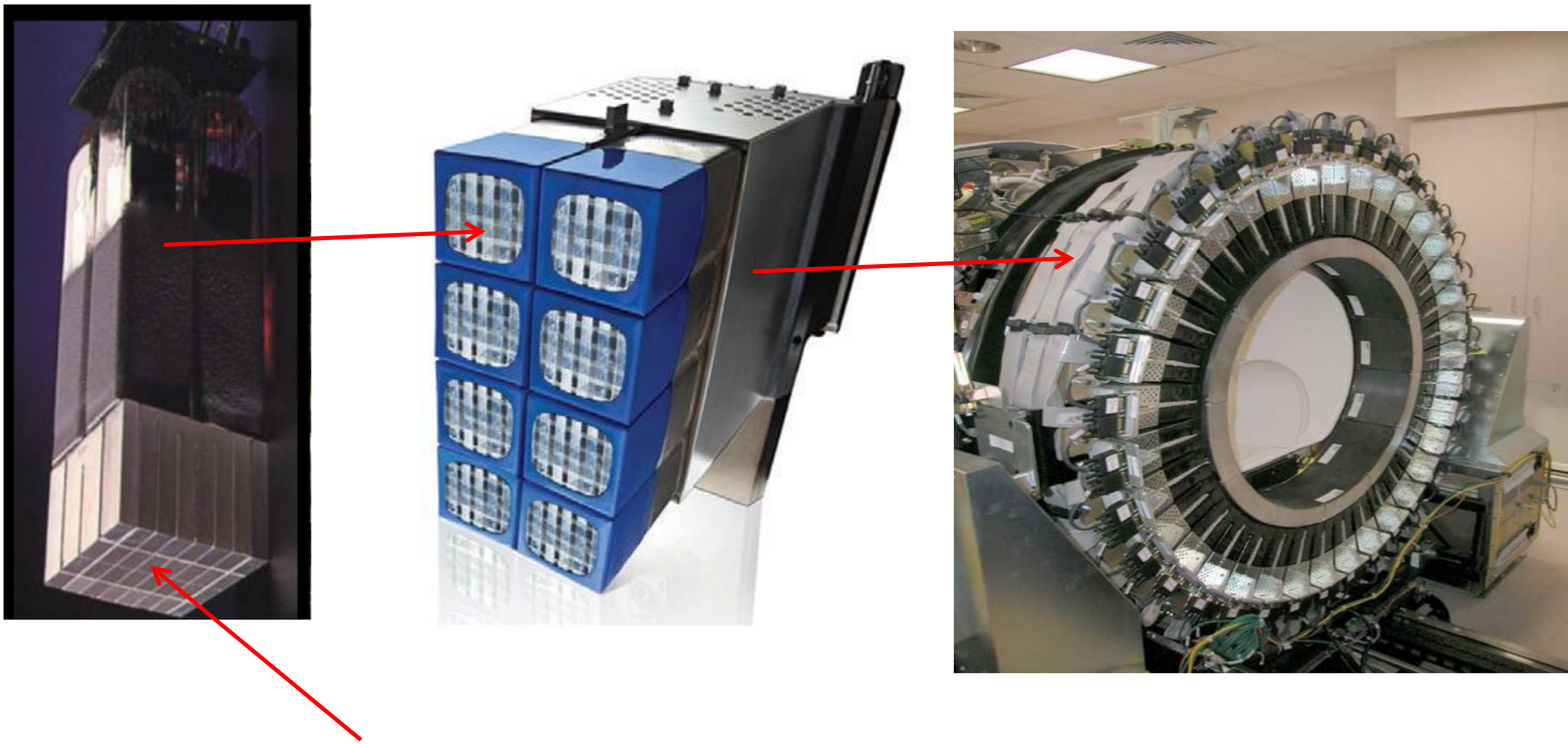
- Decay Time, Attenuation Length, and Energy Resolution: "**Less is Better**"
- Light Output: "**More is Better**"

PET detector design

Modules

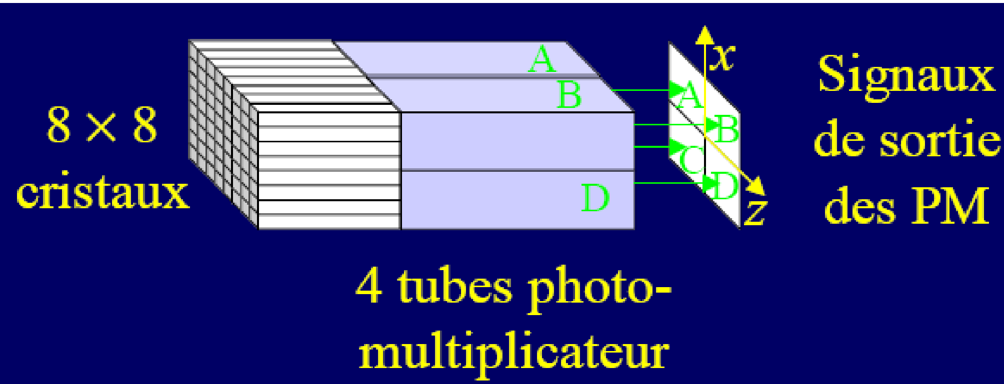
Bloc

PET ring



Detector elements (scintillators)

Block detector positioning

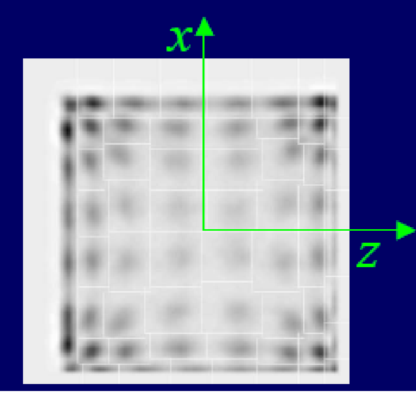


1) Calcul du centre de gravité de la distribution de lumière

$$x = \frac{(A+B)-(C+D)}{A+B+C+D}$$

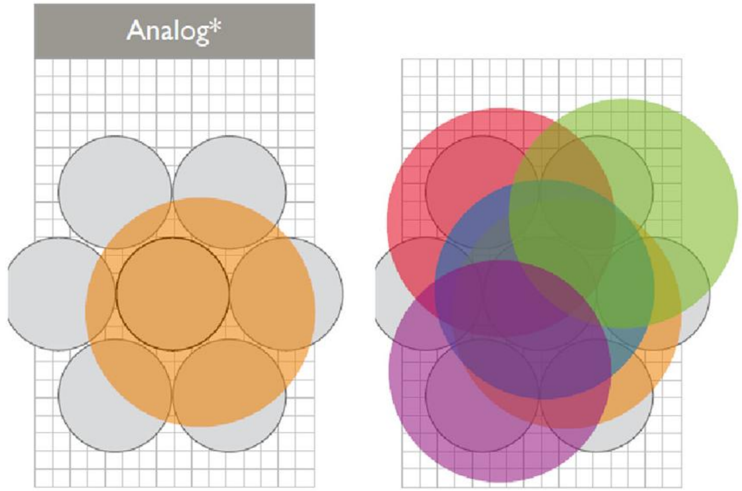
$$z = \frac{(B+D)-(A+C)}{A+B+C+D}$$

2) Identification du cristal à partir d'une image 2D mémorisée (look-up table, LUT)

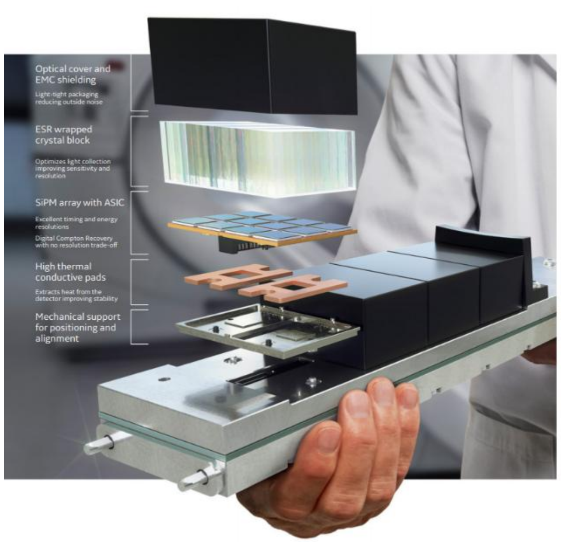


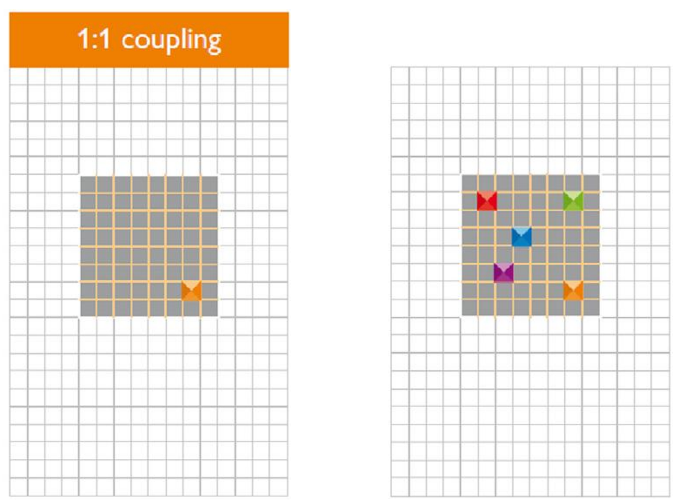
Digital vs. Analog PET technology

Analog PMT



Digital SiPM

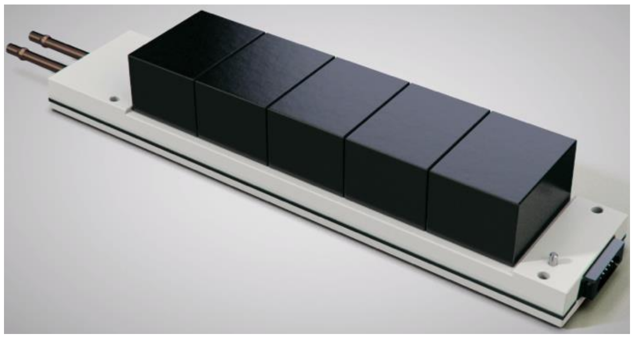


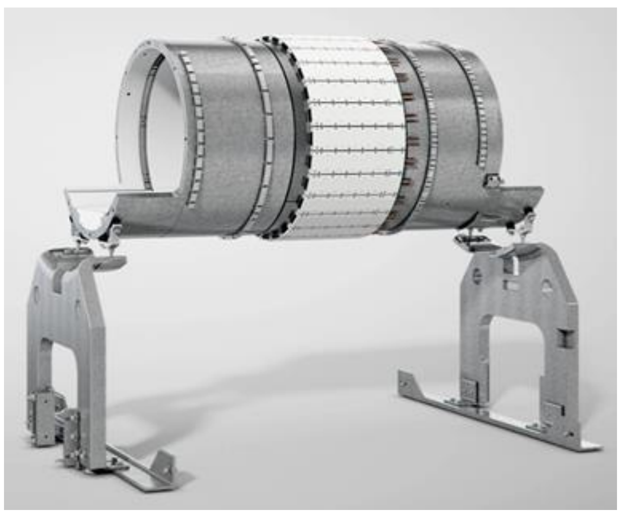


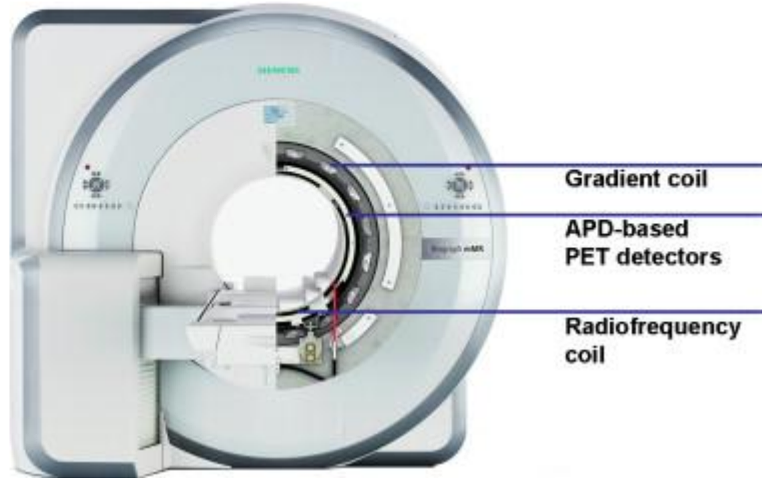
- ➢ Improved space resolution
- ➢ Improved signal localization
- ➢ Reduced signal Pile-up
- ➢ Higher count rate achievable

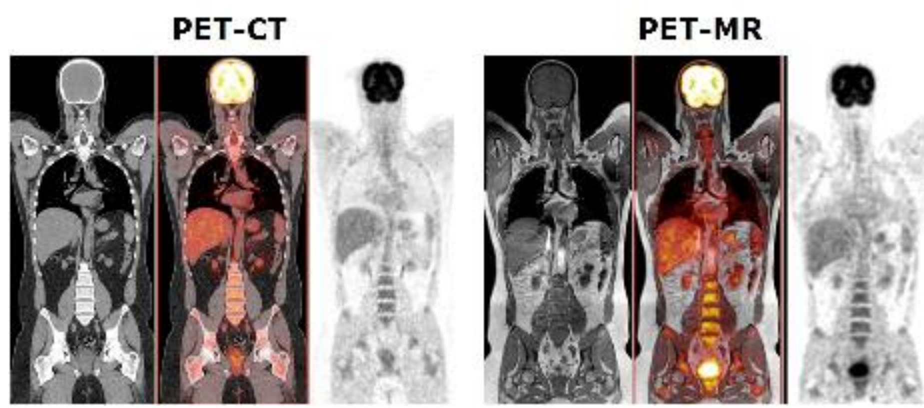
Source: Ge and Philips

Digital PET technology is compatible with strong magnetic fields









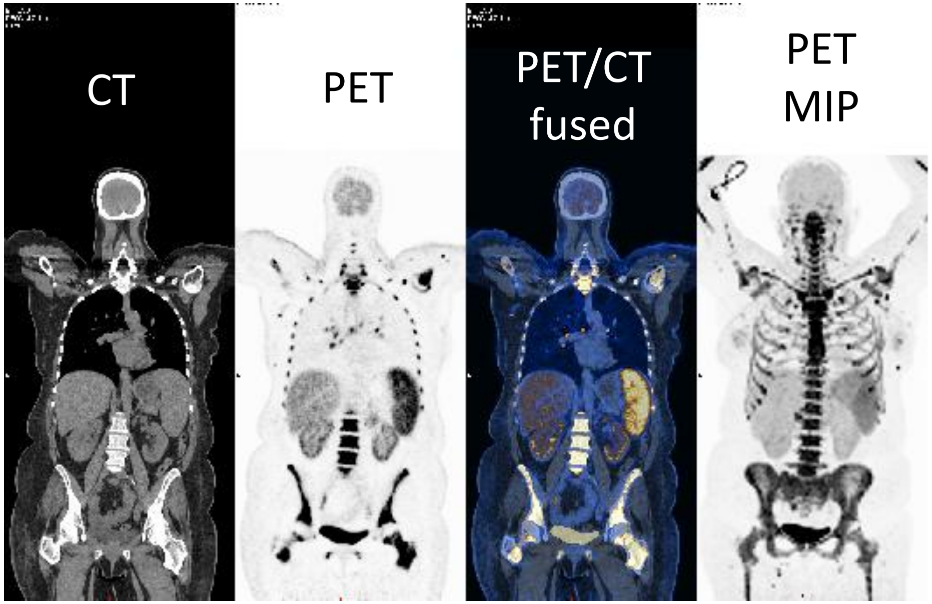
Source: GE and Siemens

Table – Comparison of the Philips Ingenuity TF, GE Discovery 710, Biograph mCT Flow and the new Philips digital PET/CT Vereos.

Model Product Name	Ingenuity TF	Discovery 710	Biograph mCT Flow	Vereos
Patient port [cm]	70 OpenView	70	78	70
Patient scan range [cm]	190	200	195	190
Maximum patient weight [kg (lb)]	195 (430)	226 (500)	226 (500)	195 (430)
Acquisition modes	3D S&S	3D S&S	3D S&S, continuous	3D S&S
Number of image planes	45 or 90	47	109	72
Plane spacing [mm]	2 or 4	3.27	2	1, 2, or 4
Crystal size [mm]	4 × 4 × 22	4.2 × 6.3 × 25	4 × 4 × 20	4 × 4 × 22
Number of crystals	28,336	13,824	32,448	23,040
Number of PMTs	420	256	768	23,040 SiPMs
Physical axial FOV [cm]	18	15.7	21.8	16.3
Detector material	LYSO	LYSO	LSO	LYSO
System sensitivity 3D, [%]*	0.74	0.75	0.95	>1.0
Trans axial resolution @ 1 cm [mm]*	4.7	4.9	4.4	4.0
Trans axial resolution @ 10 cm [mm]*	5.2	5.5	4.9	4.5
Axial resolution @ 1 cm [mm]*	4.7	5.6	4.5	4.0
Axial resolution @ 10 cm [mm]*	5.2	6.3	5.9	4.5
Peak NECR [kcps]	120 @19 kBq/ml	130 @29.5 kBq/ml	175 @28 kBq/ml	400 @30 kBq/ml
Time-of-flight resolution [picoseconds]	591	544	540	307
Time-of-flight localization [cm]	8.9	8.2	8.1	4.6
Coincidence window [nanoseconds]	4.5	4.9	4.1	1.5

The sensitivity, NECR (noise equivalent count rate), coincidence window and TOF resolution are higher for the digital PET/CT digital PET/CT.
Abbreviations: FOV, field-of-view; PMT, photomultiplier tubes; NECR, noise equivalent count rate; kcps, kilocounts per second; kBq/ml, kiloBecquerel/milliliter; S&S, step and shoot.
* NEMA 2001.

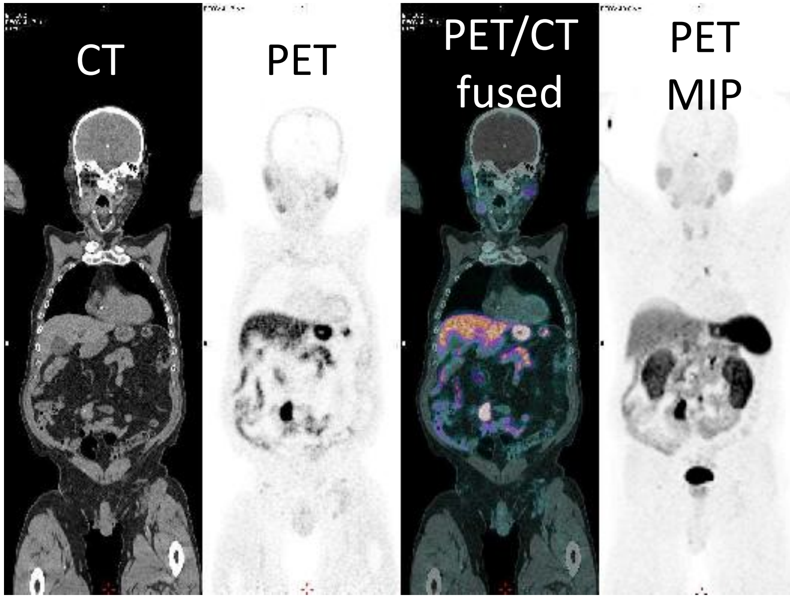
PET: some clinical application-1



F-18 FDG PET/CT of Lymphoma

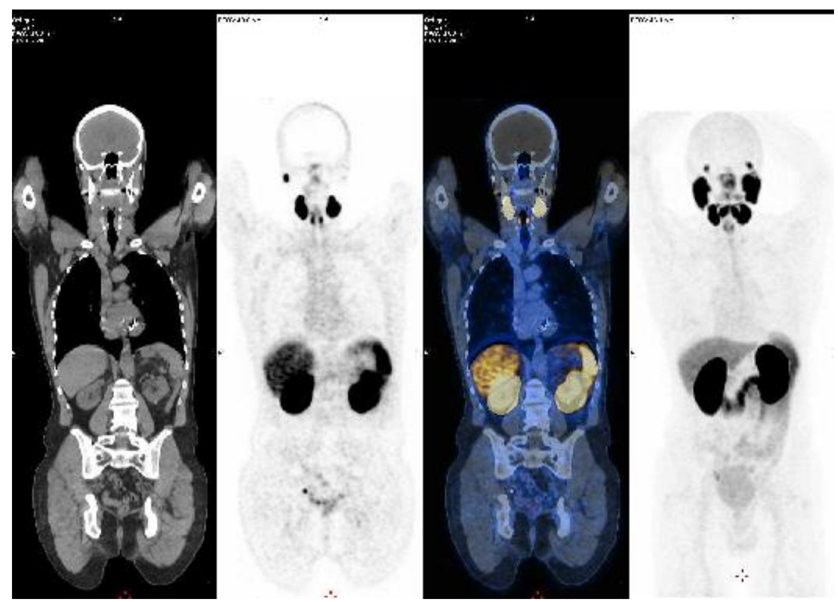
F-18 FDG PET/CT of ovarian carcinoma

PET: some clinical application-2

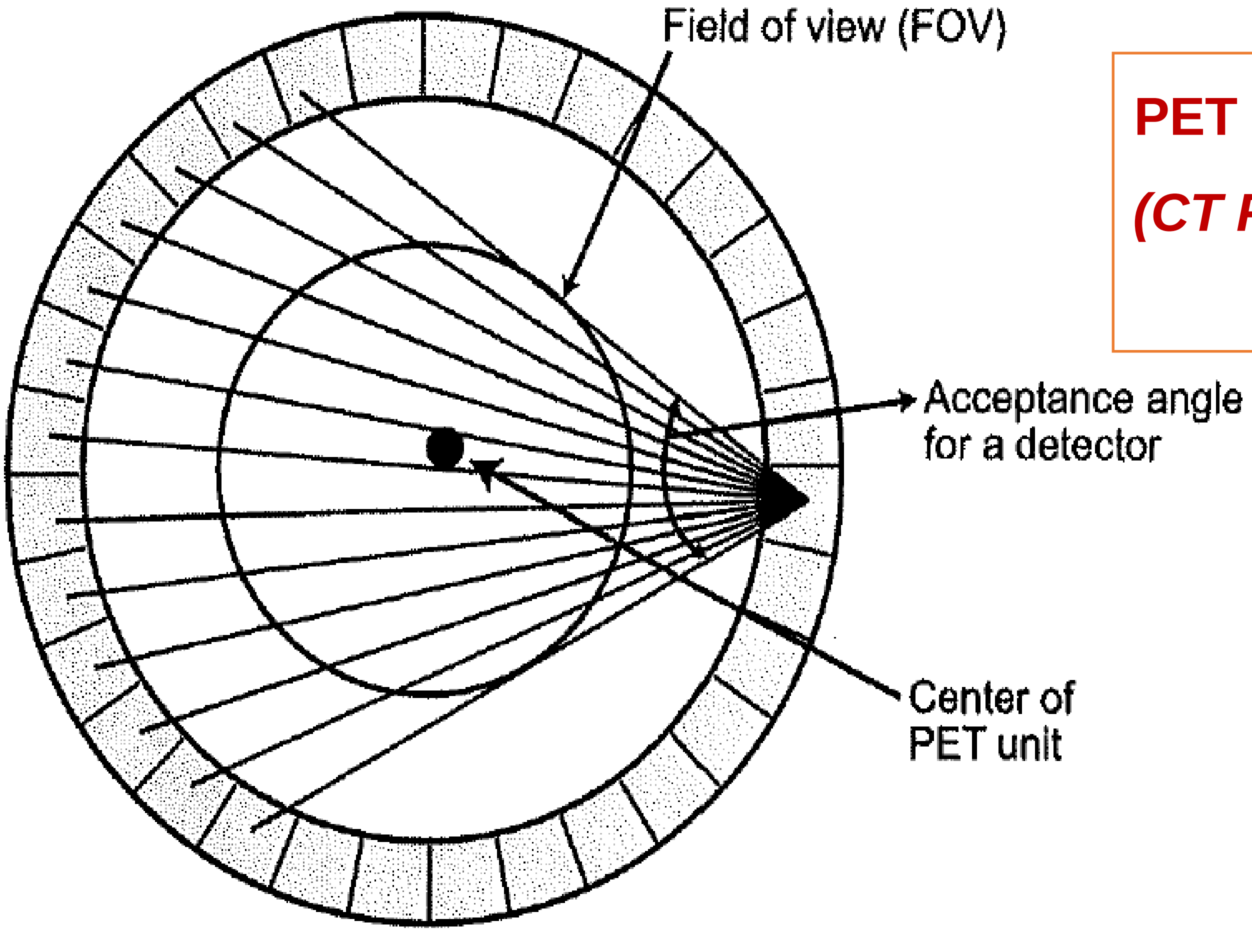


Ga-68 DOTATATE PET/CT of Neuroendocrine Tumour

Ga-68 PSMA PET/CT of prostatic carcinoma



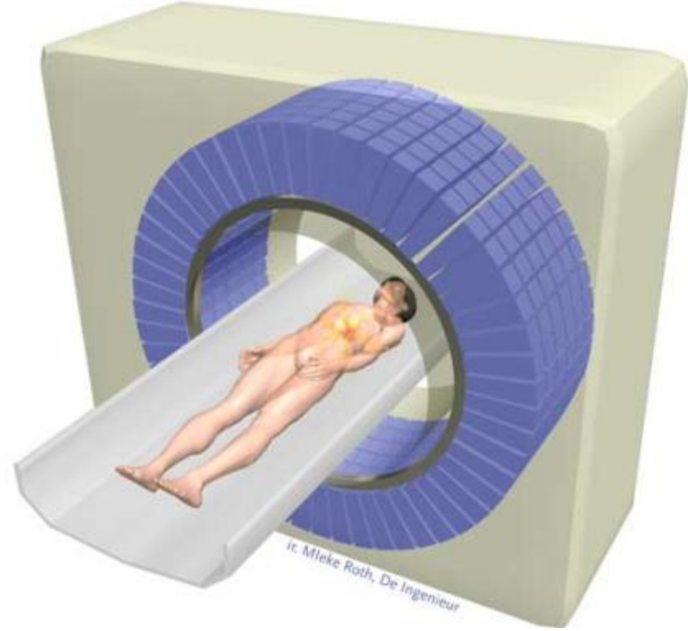
PET: Field of view (FOV)

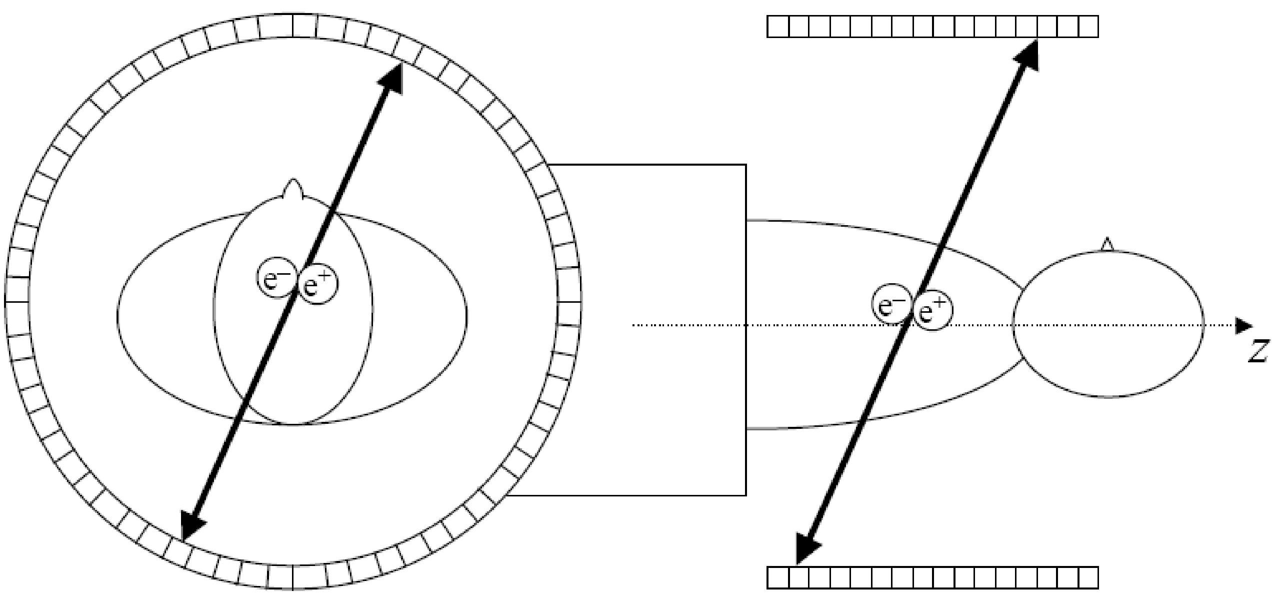


PET FOV : 60-70 cm

(CT FOV : 45-50 cm)

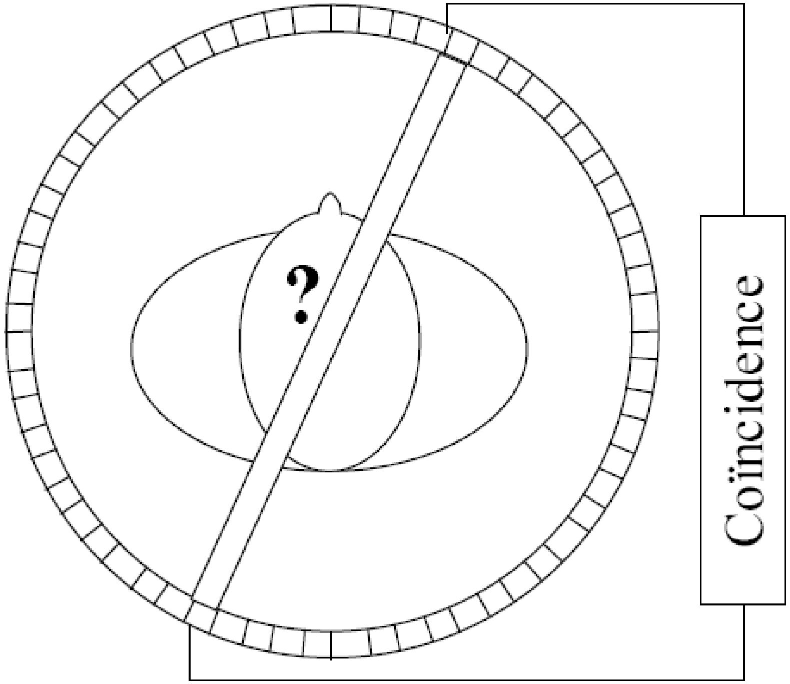
2D vs. 3D data acquisition





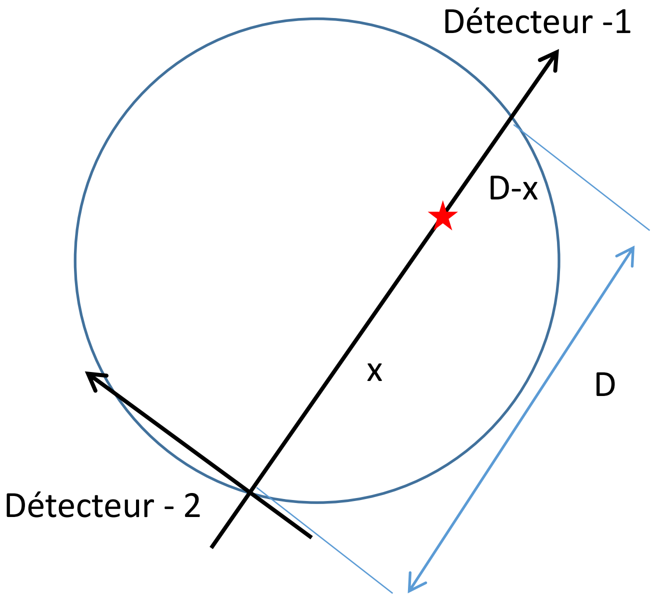
Spatial localization of annihilation events

2-511keV photon in coincidence → 1 Line (tube) Of Response (LOR)



- **Goal**
  - Recover the exact position of the annihilation event
- **Problem**
  - We have not information about the place along the LOR where the annihilation happened

Direct localization using the Time of Flight (TOF) information



Measured detection time in D1 and D2

$$T_1 = \frac{D-x}{c} \qquad T_2 = \frac{x}{c}$$

$$\Delta T = T_2 - T_1 = \frac{x}{c} - \frac{D-x}{c} = \frac{2x-D}{c}$$

If ΔT = 0 → x =D/2

If ΔT = 0.2ns and D = 0.8m→ x =43cm

$$\delta T = \frac{\delta x}{c}$$

Error on time estimation →
Error in annihilation position estimation

$$\Delta x = \frac{c\Delta t}{2} \Leftrightarrow \Delta t = \frac{2\Delta x}{c}$$

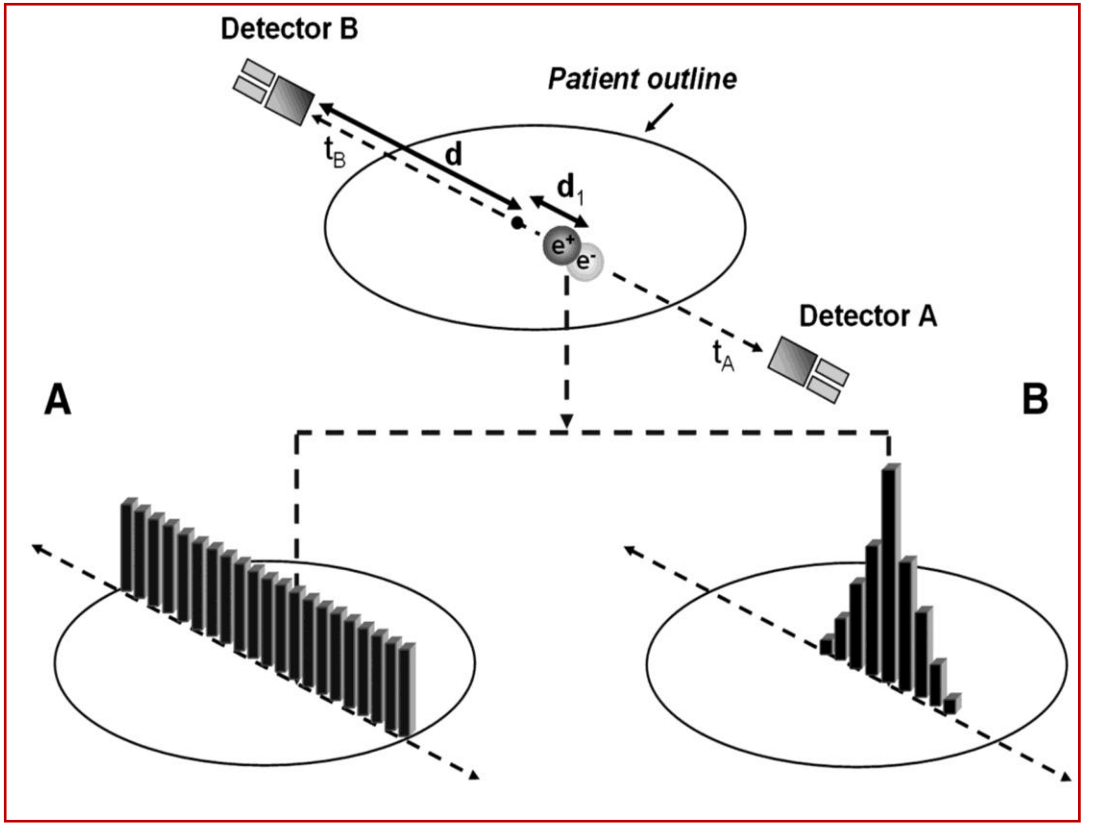
Present generation TOF time resolution δT = 300 ps → δx = (C x 300 ps)/2 = 4.5cm

Present TOF performances are not sufficient for pure TOF image reconstruction

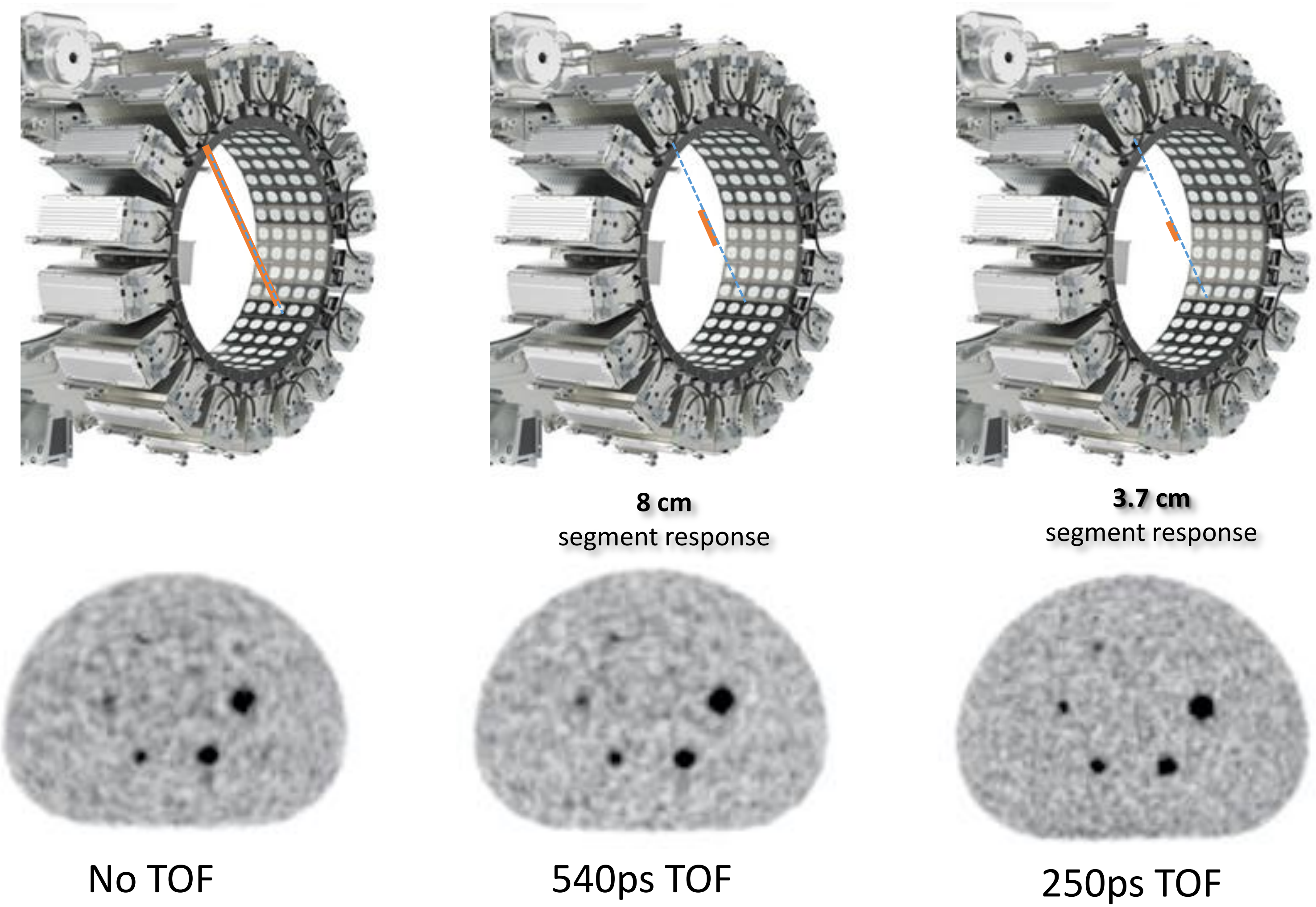
Direct localization using the Time of Flight (TOF) information

- Time resolution achievable : 300 ps
  - ➢ Achievable spatial resolution : 45 mm

- Direct (full) TOF image reconstruction still not possible
  - ➢ Useful information to better conditioning the tomographic reconstruction problem

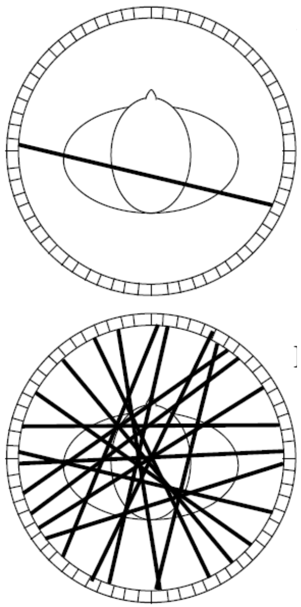


Advancement in TOF PET

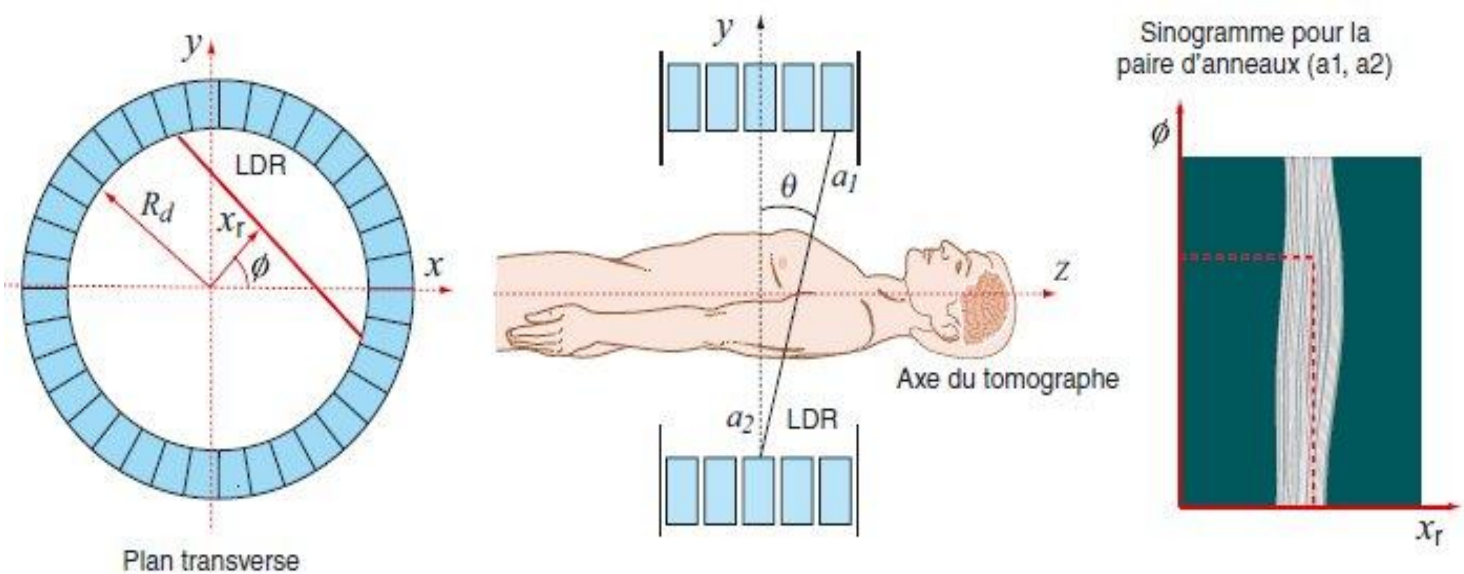


- Improved image contrast
- Improved lesion detectability

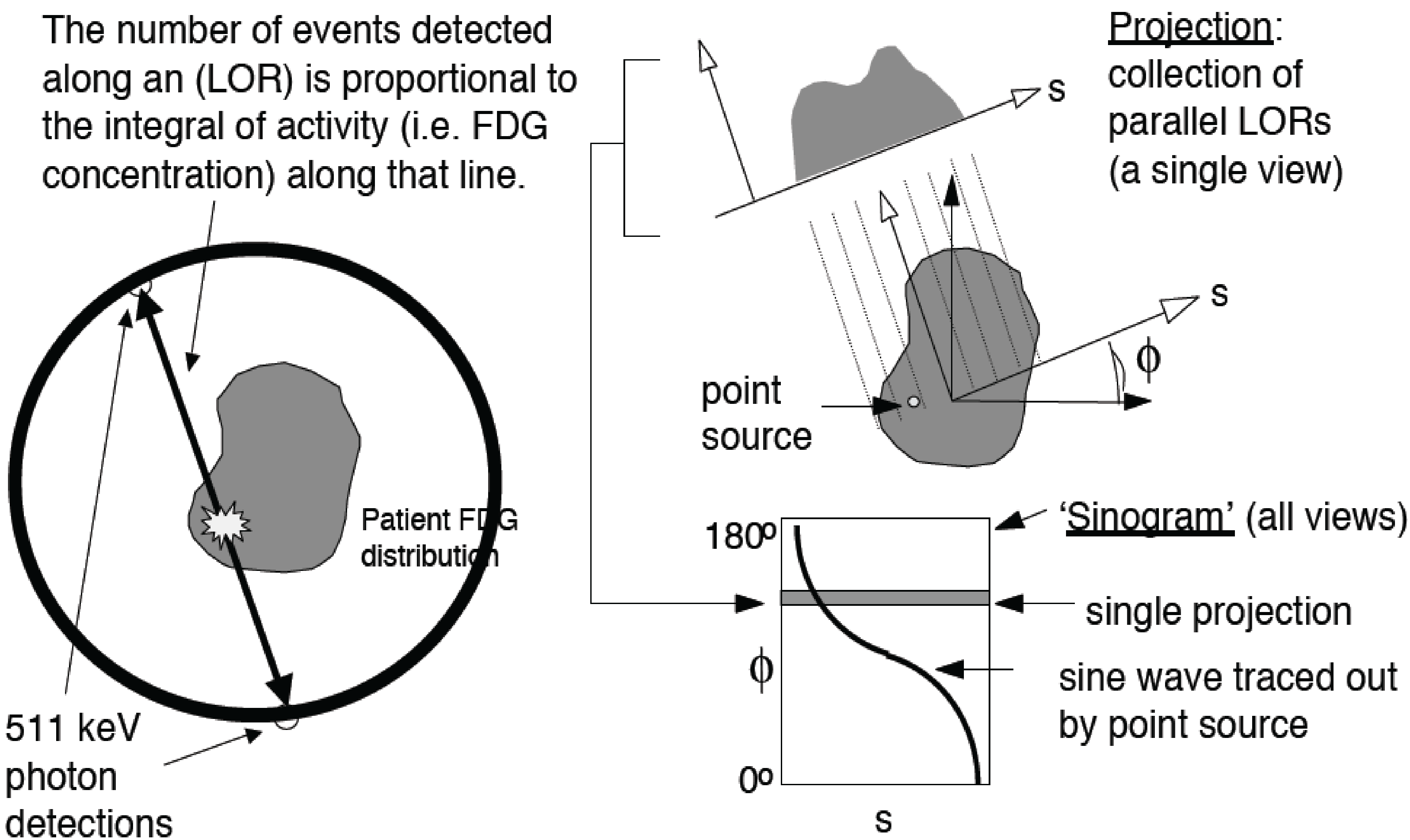
Tomographic reconstruction: the sinogram



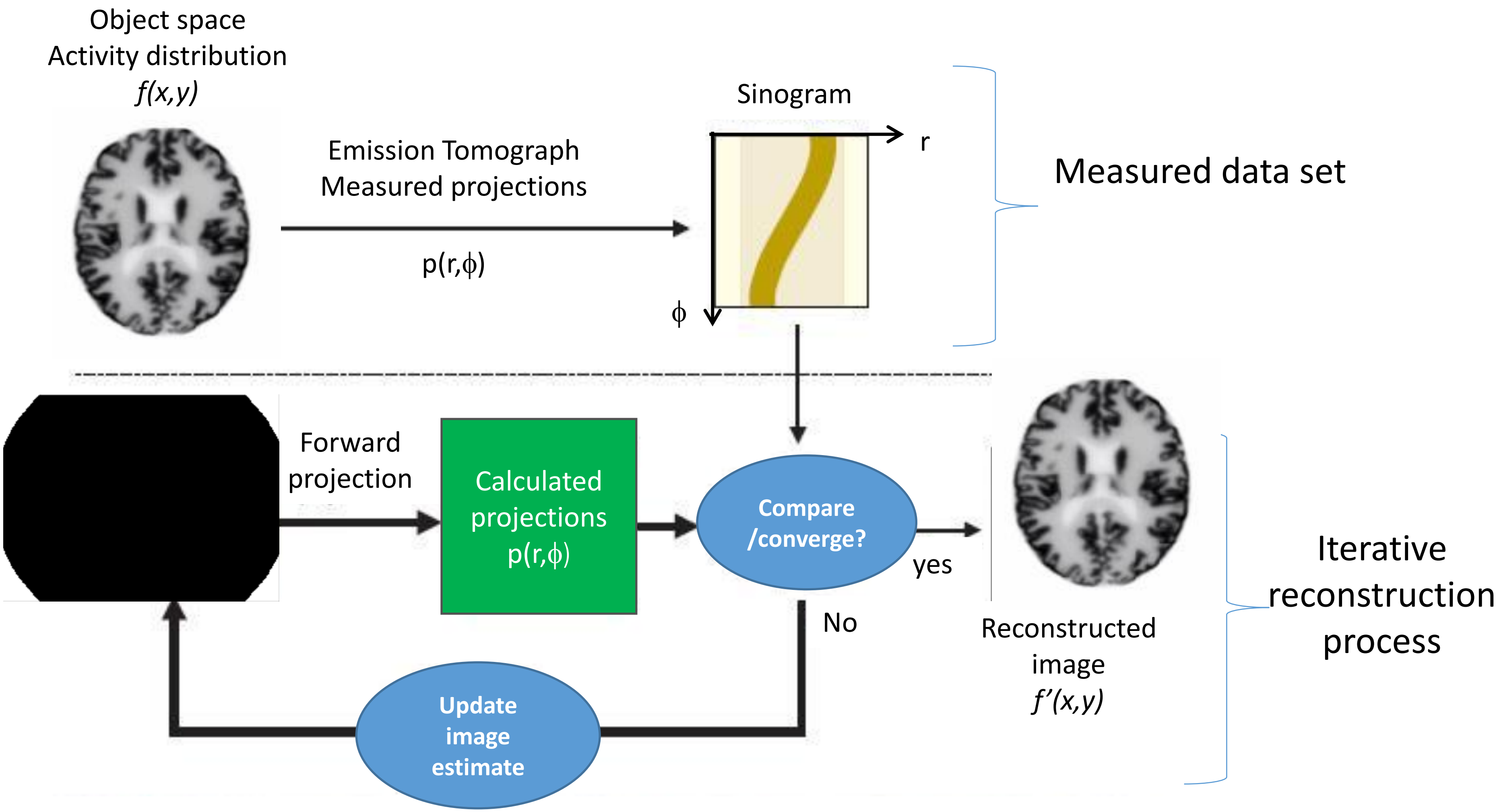
- 1 LOR (one detected coincidence)
  - The annihilation happened somewhere along the LOR. No direct localisaton
- Many LORs
  - The activity distribution cab be obtained from tomographic reconstruction techniques.



Sinogram representation



Example of Iterative Reconstruction



Animations from:
Floris HP van Velden, PhD
EANM Milan 2012

Iterative reconstruction

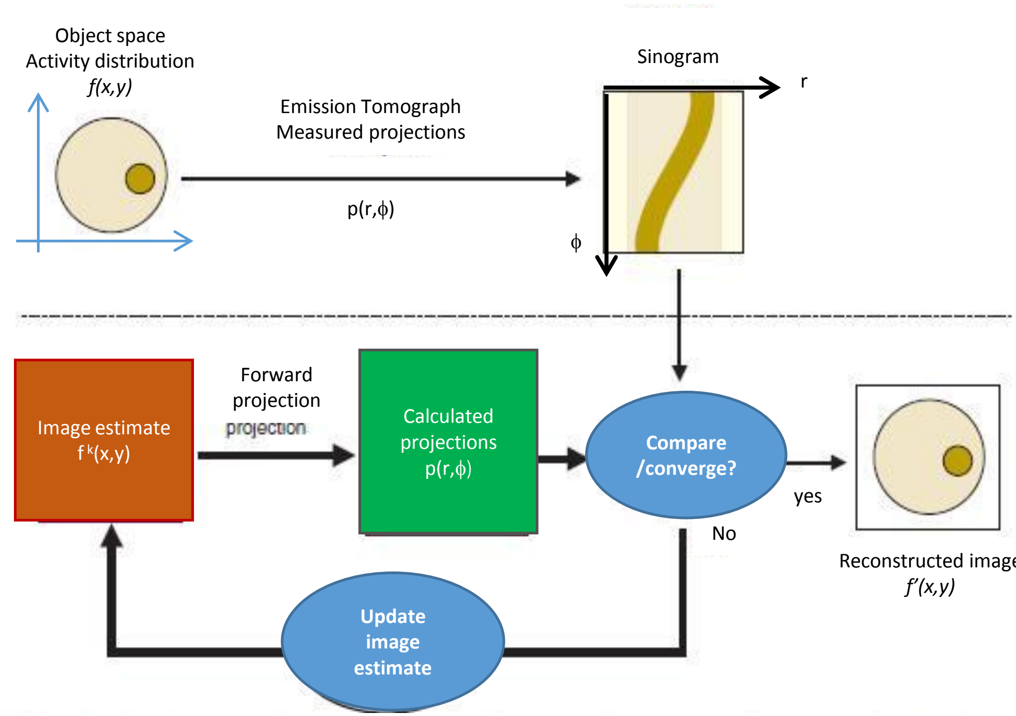
Key points:

Expectation maximization cost function *(convergence criterion)*

Method to upgrade the image estimate

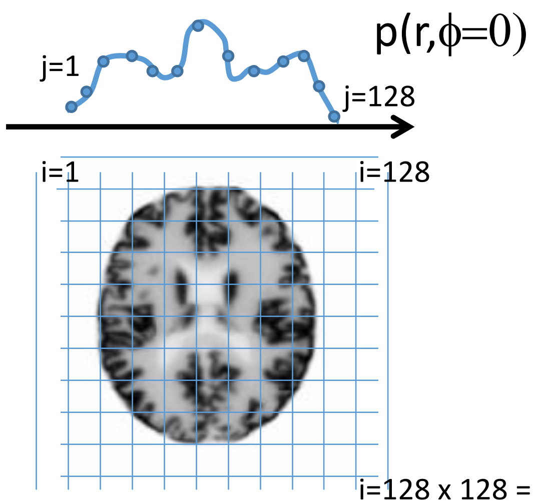
Account for:

- *Detector PSF response,*
- *TOF information*
- *Attenuation*
- *Scattering*



Quantitative more accurate than FBP

All image pixels (i) have a finite probability ($M_{i,j}$) to contribute to signal intensity into projection data (p_j)



i=128 x 128 = 16348

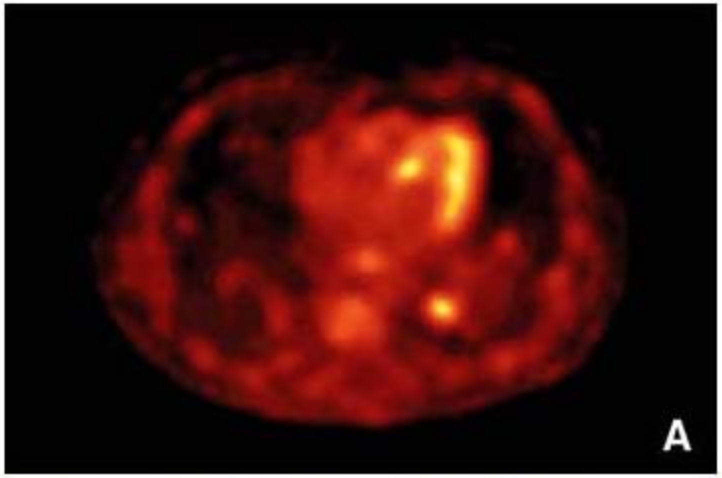
$$p_j = \sum_{i=1}^{n} M_{i,j} f_i$$

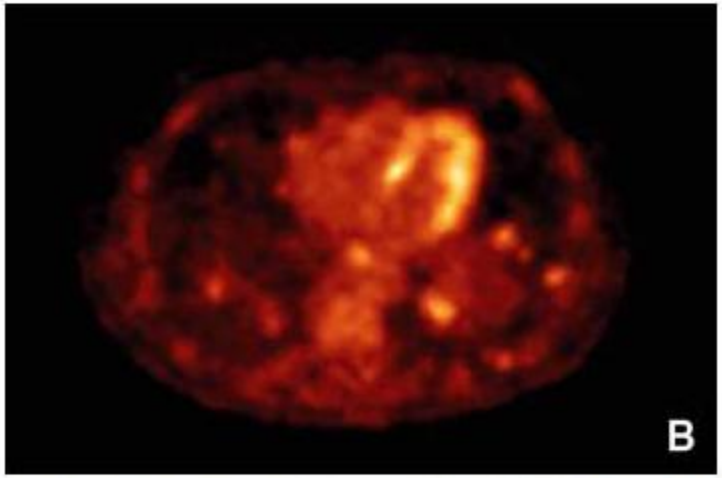
$$f_i^{k+1} = \frac{f_i^k}{\sum_{j=1}^{m} M_{i,j}} \times \sum_{j=1}^{m} M_{i,j} \frac{p_j}{\sum_{i'=1}^{n} M_{i',j} f_{i'}^k}$$

If 120 angular projections → 120 x 128 = 15360 projection data → M : (16348 x 15360)

TOF improvement in image quality and quantification

Improved SNR

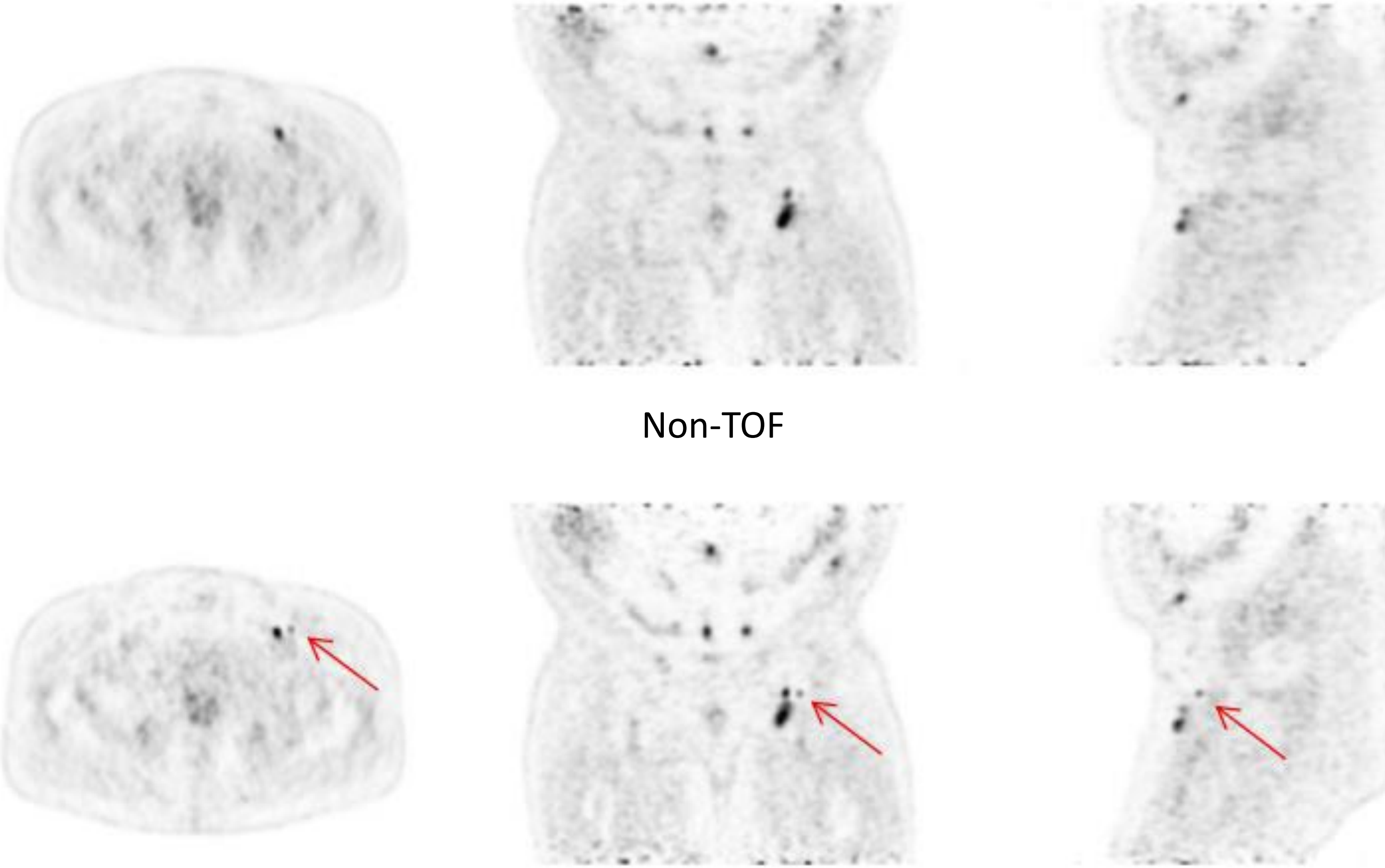




PET reconstruction without TOF

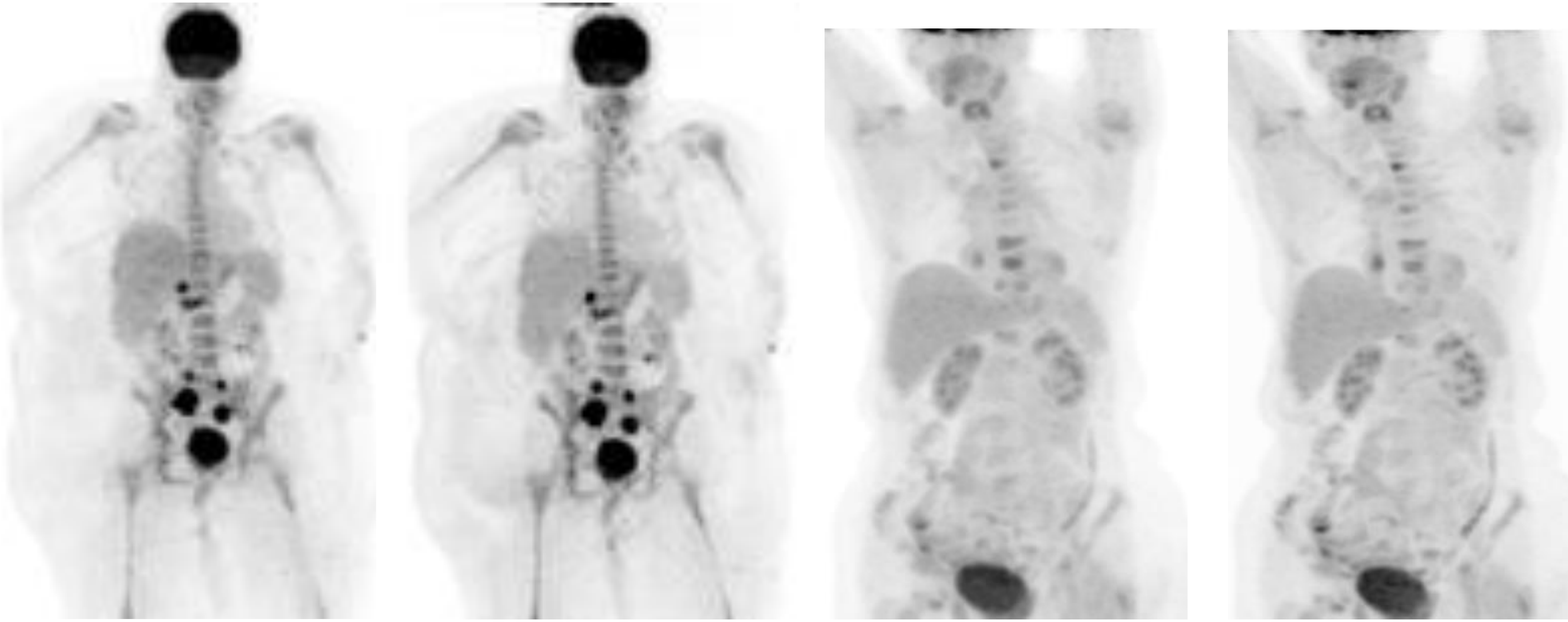
Reconstruction with TOF

TOF Improved pelvic nodule visualization

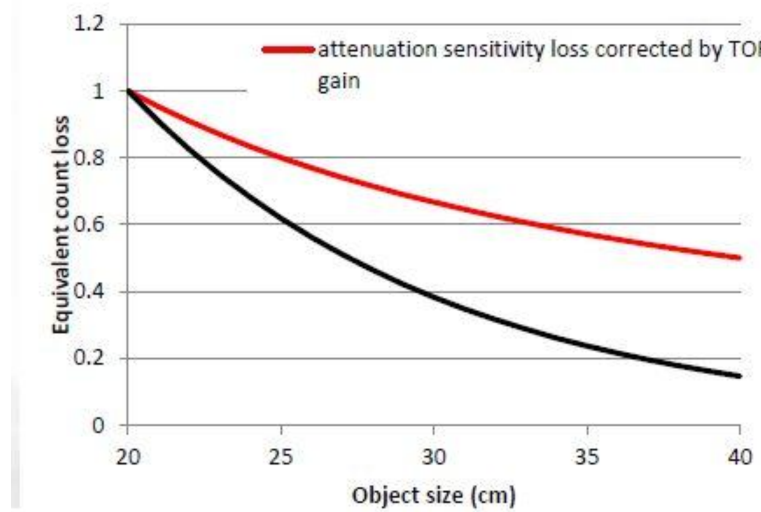


Non-TOF

TOF



Gain based on: $Sensitivity_{gain} \propto 2 \times \frac{D(cm)}{\Delta t(ns) \times c}$

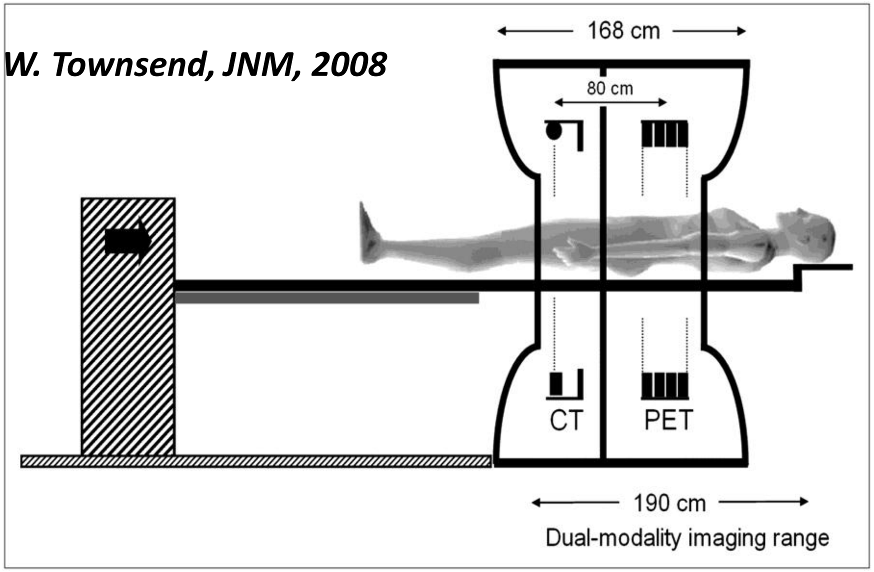


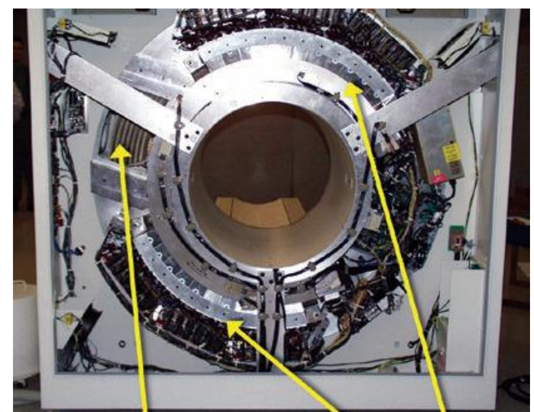
Body attenuation greatly reduces counts. As size increases, counts are reduced exponentially. ToF gain is greater for large patients as it partially compensates for the lower quality of large patients

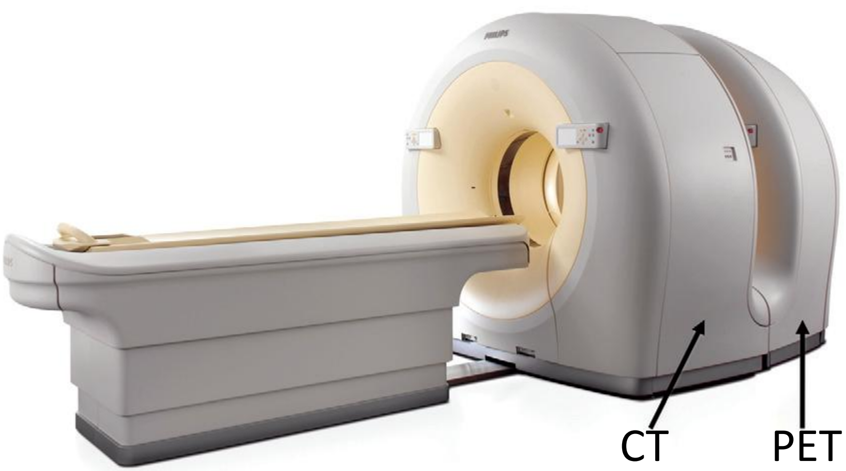
Dual modality PET-CT

- **Goal** : improve activity localization and implement attenuation correction (auto-registration of anatomic CT and functional PET)

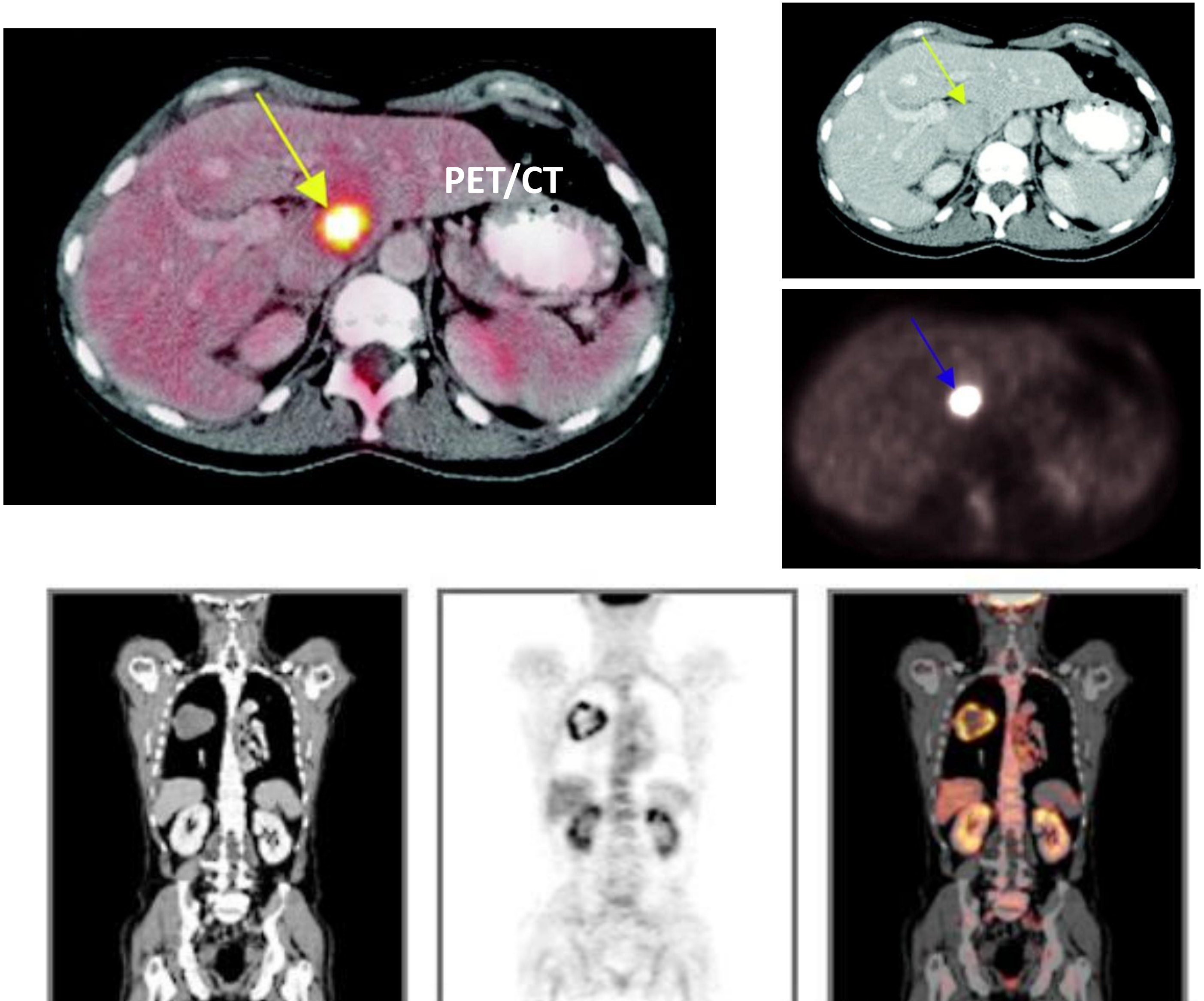
D. W. Townsend, JNM, 2008







Dual modality PET-CT



Commercially available digital SiPM PET/CT and comparison with an analog PMT-based PET/CT

Philips Vereos

TOF : 310 ps
Axial extension :164 mm
NEMA sensitivity : 5.7 kcps/MBq
TOF eff. Sensitivity : 22kcps/MBq

GE Discovery MI

TOF : 370 ps
Axial ring extension: 200mm
NEMA Sensitivity : 13.5 kps/MBq
TOF eff. sensitivity : 48 kcps/MBq

Siemens Biograph Vision

TOF : 250 ps
Axial ring extension: 248mm
NEMA sensitivity : 15 kcps/MBq
TOF effe. Sensitivity : 84kcps/MBq

GE Discovery 690

TOF : 540 ps
Axial ring extension: 153 mm
NEMA Sensitivity : 7.5 kBq/MBq
TOF eff. Sensitivity : 19 kcps/MBq

Spatial resolution (F-18) ~4mm

Clinical acquisition/reconstruction parameters

Philips Vereos

Admin. A : 2 MBq/kg
Time per bed position: 90 s
TAP = 3 min.MBq/kg
OSEM param: 3it × 15ss TOF+PSF
Image smoothing: NONE
Image matrix: 288×288
Pixel size: 2×2×2 mm

GE Discovery MI

Admin. A : 2.5 MBq/kg
Time per bed position: 90 s
TAP = 3.75 min.MBq/kg
OSEM param: 3it × 16ss TOF+PSF
Image smoothing: Gaussian 6.4mm
Image matrix: 256×256
Pixel size: 2.73×2.73×2.79 mm

Siemens Biograph Vision

Admin. A : 2 MBq/kg
Time per bed position: 120 s
TAP = 4 min.MBq/kg
OSEM param: 3it × 5ss TOF+PSF
Image smoothing: NONE
Image matrix: 440×440
Pixel size: 1.65×1.65×2 mm

GE Discovery 690

Admin. A : 3.5 MBq/kg
Time per bed position: 90 s
TAP = 5.25 min.MBq/kg
OSEM param: 3it × 16ss TOF+PSF
Image smoothing: Gaussian 5mm
Image matrix: 256×256
Pixel size: 2.73×2.73×3.27 mm

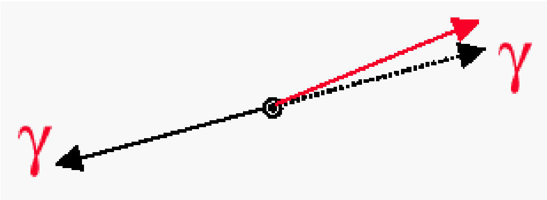
Which parameter does not influence the spatial resolution in PET?

1. The photon energy
2. Positron energy
3. Detector size (ring diameter)
4. Reconstruction algorithm

Spatial resolution

$$R_t = K_r \cdot \sqrt{R_i^2 + R_p^2 + R_a^2 + R_l^2}$$

- R_i is related to the detector width (w)
  - from w/2 (center) to w (detector), 2 – 4 mm
- R_p is related to the positron range
- 0.2 mm for ^{18}F and 2.6 mm for ^{82}Rb
- R_a is related to the γ non-colinearity
  - $\pm$ 0.25° deviation from 180°
  - 1.8 mm for a 80-cm PET scanners



- R_l is related to the localization of detector
  - (use of block detectors instead of single detectors)
  - 2.2 mm for BGO (less for LSO)
- K_r is a factor related to the reconstruction technique (1.2 to 1.5)
- R_t at the center of the FOV : 5 mm for ^{18}F

Detection efficiency

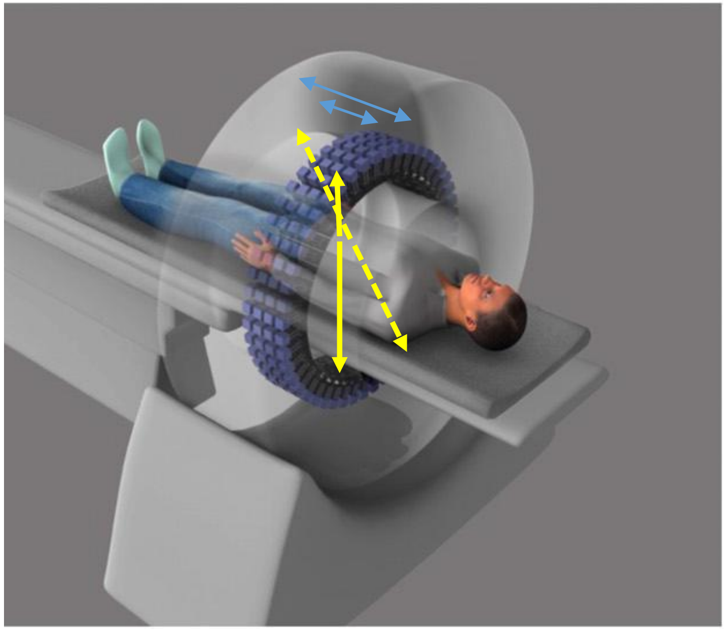
$$S = \frac{A}{4\pi r^2} \cdot \varepsilon^2 [cps/MBq]$$

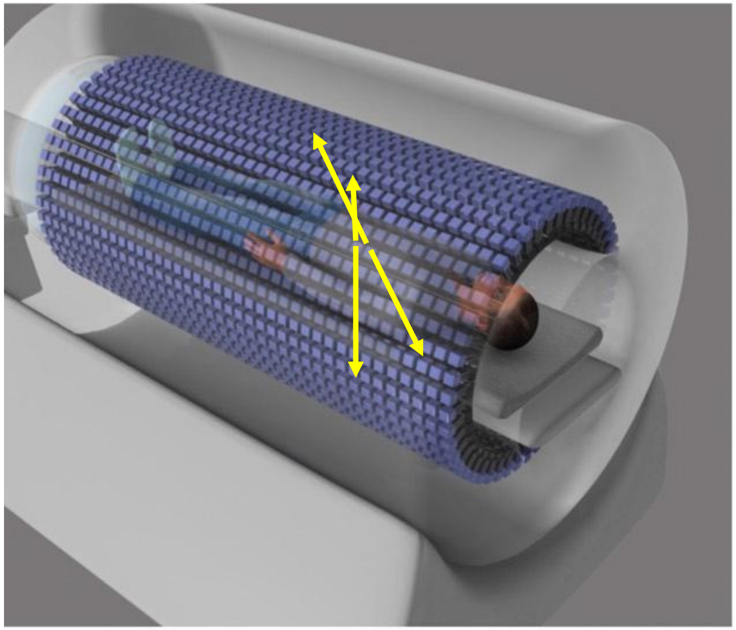
- A = detector area seen by a point source to be imaged
- ε = 1-exp(-μ(Eγ=511keV,det)x) detector's efficiency
- μ = linear attenuation coefficient of 511 kev photons in the detector
- x= thickness of the detector
- r = radius of the detector ring

- S = 0.2 – 0.5 % for 2D PET and 1-10% for 3D PET
- *(S = 0.01-0.03% for SPECT)*

- Manufacturer provides volume sensitivity S_{vol} [cps/Bq/ml]

Improving systems sensitivity



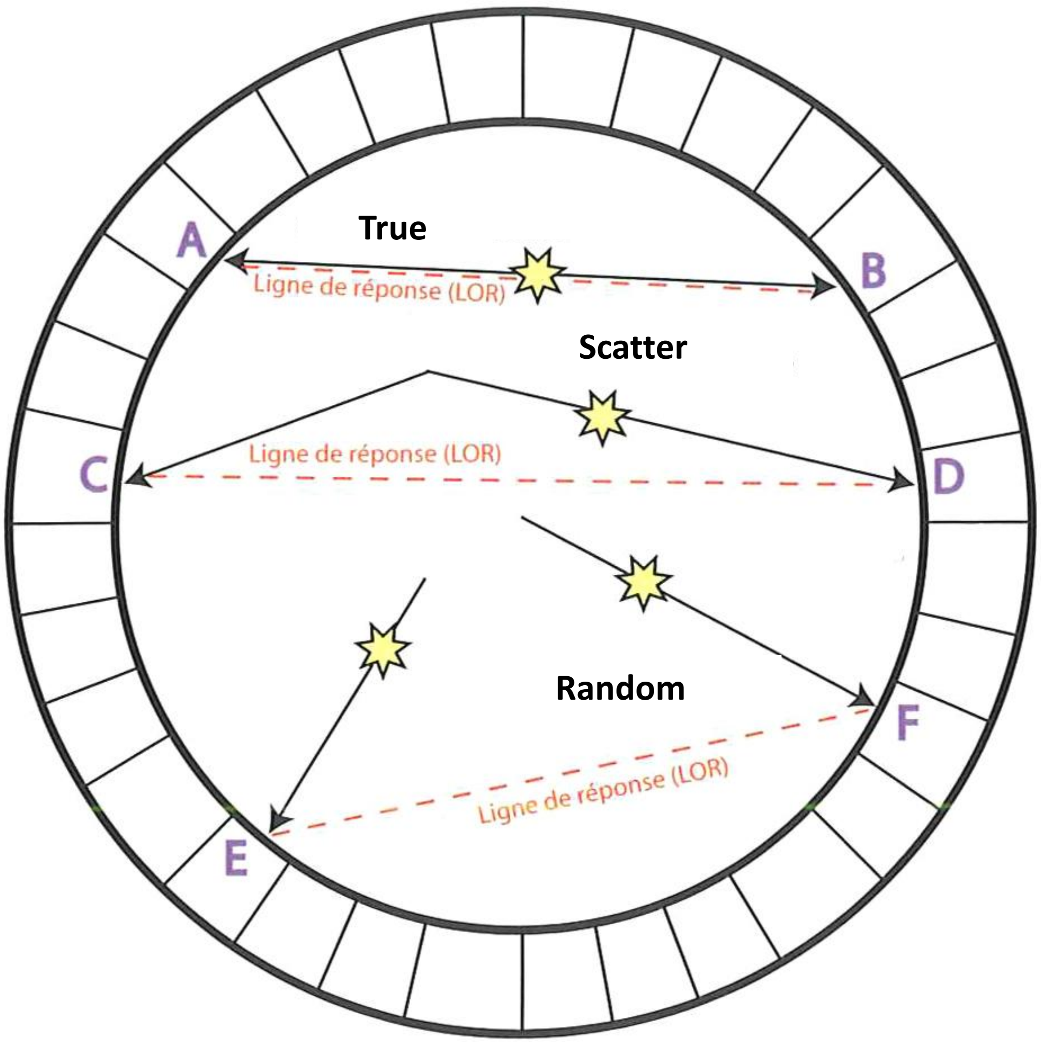


EXPLORER Total Body PET S.R.Cherry et al. J Nucl Med 2018; 59:3–12

- Increasing PET ring coverage (solid angle) — Costs
- Increase crystal thickness (increase probability of interaction) — Light collection efficiency; Depth of interaction
- Increase energy windows width around the 511keV peak — Scatter and prompt gamma pollution

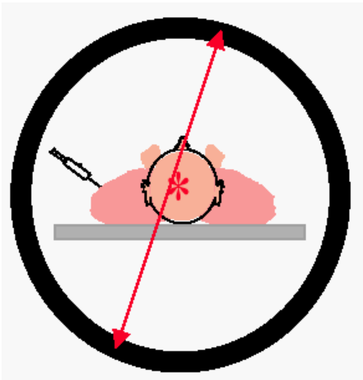
Margin for reducing patient admin. activity and thus patient dose
Margin for reducing exam duration with consequent improved patient comfort

Coincidence Types in PET

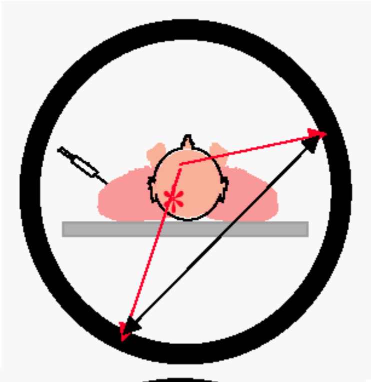


Coincidence types in PET

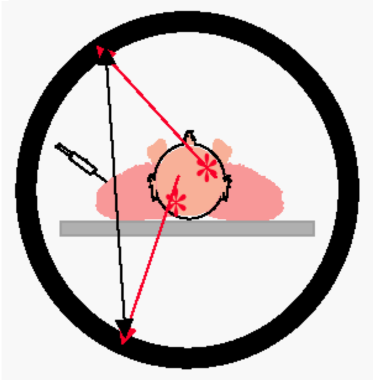
- **True coincidence**
  - ☞ Correct localization long the LOR
  - ☞ Useful for image reconstruction



- **Scatter coincidence (Compton)**
  - ☞ Mislocalization
  - ☞ Contrast reduction
  - ☞ Quantitative bias

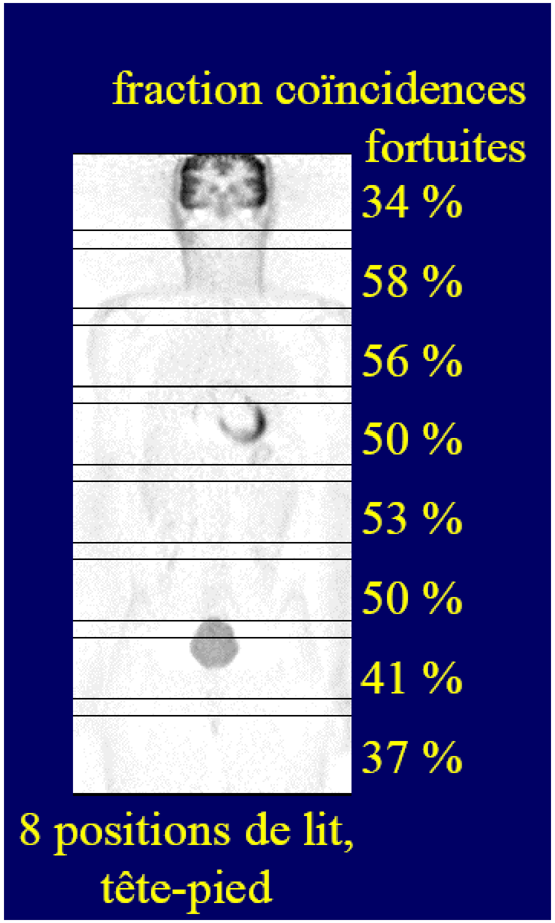


- **Random coincidence**
  - ☞ Mislocalization
  - ☞ Important component → count rate saturation
  - ☞ Quantitative bias



Random coincidences

Spatial activity distribution

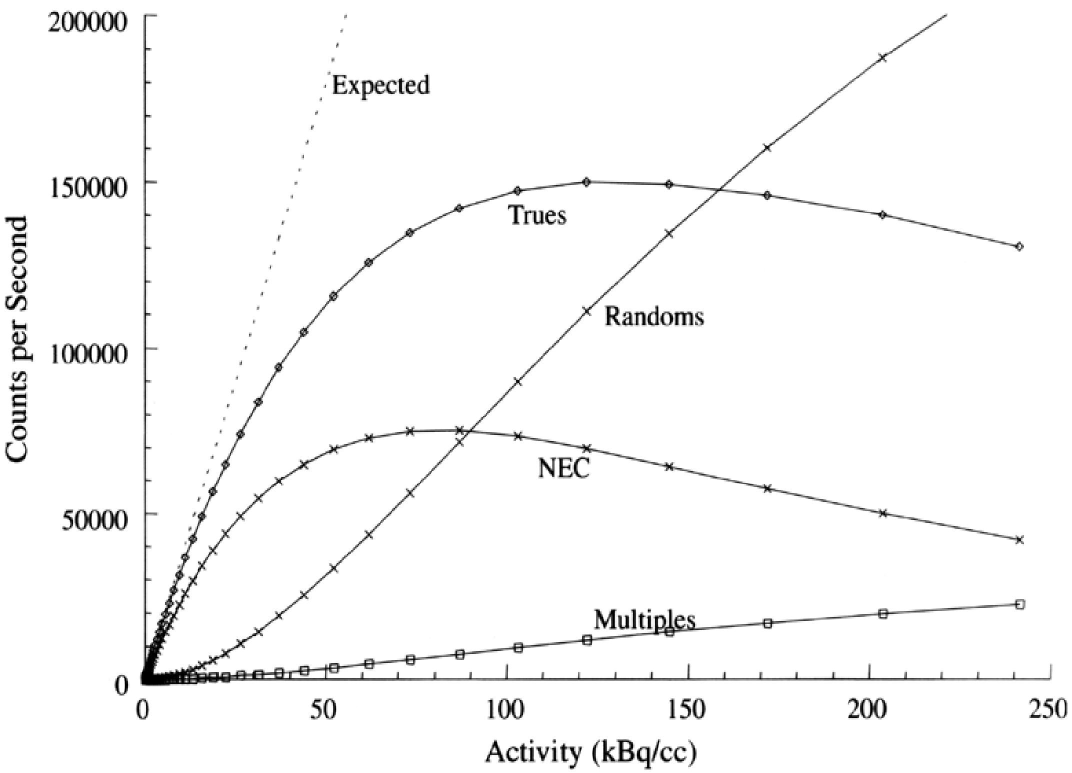


- The number of true coincidences is proportional to the activity
- The number of random coincidences:
  - $N_{random} = 2\tau S_1 S_2$
  - S_1 and S_2 : number of events in a couple of detectors (1 and 2) in coincidence.
  - → $N_{random} \sim A^2$
  - **$\Rightarrow N_{random} / N_{true} \sim A$**

Noise equivalent count (NEC)

$$NEC = \frac{N_{trues}^2}{N_{trues} + N_{scatters} + N_{randoms}}$$

(NEC proportional to SNR)



$$R_{true} = A\varepsilon^2 g_\Omega e^{-\mu T} \propto A$$

$$R_{random} = \tau \times R_{\text{single-1}} \times R_{\text{single-1}} \propto \tau A^2$$

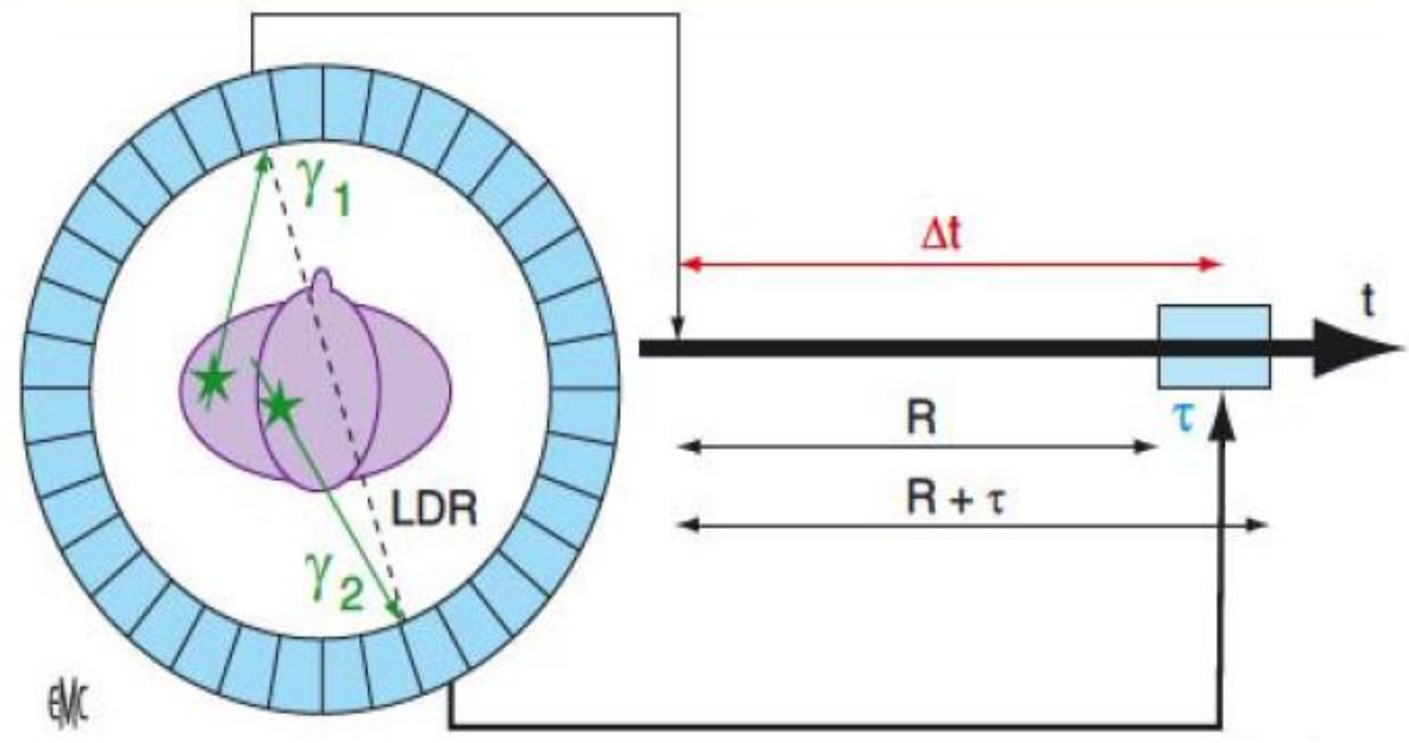
$$R_{scatter} \propto A$$

Random coincidence correction

- Reduce the time coincidence window: τ (not sufficient to obtain good results and need a fast light propagation into the detector crystal and fast signal processing)
- **Correction by a delayed coincidence window**
  - Differently from true coincidences, random coincidences are uncorrelated in time
  - Simple events are acquired in a first time window (6-12 ns)
  - Simple event detection in a ~50 ns delayed coincidence window (same width of 6-12 ns)
  - The events collected in the two time windows are artificially correlated and used to estimate the number of random coincidences
  - Real time subtraction of the ''estimated'' random coincidences from the total number of prompt coincidences. The correction (subtraction) is performed for each LOR.
  - Correction by simulation

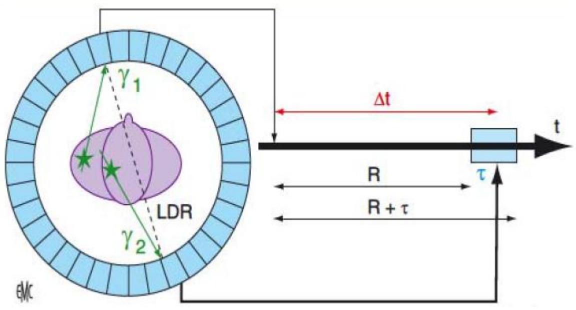
Random coincidence correction

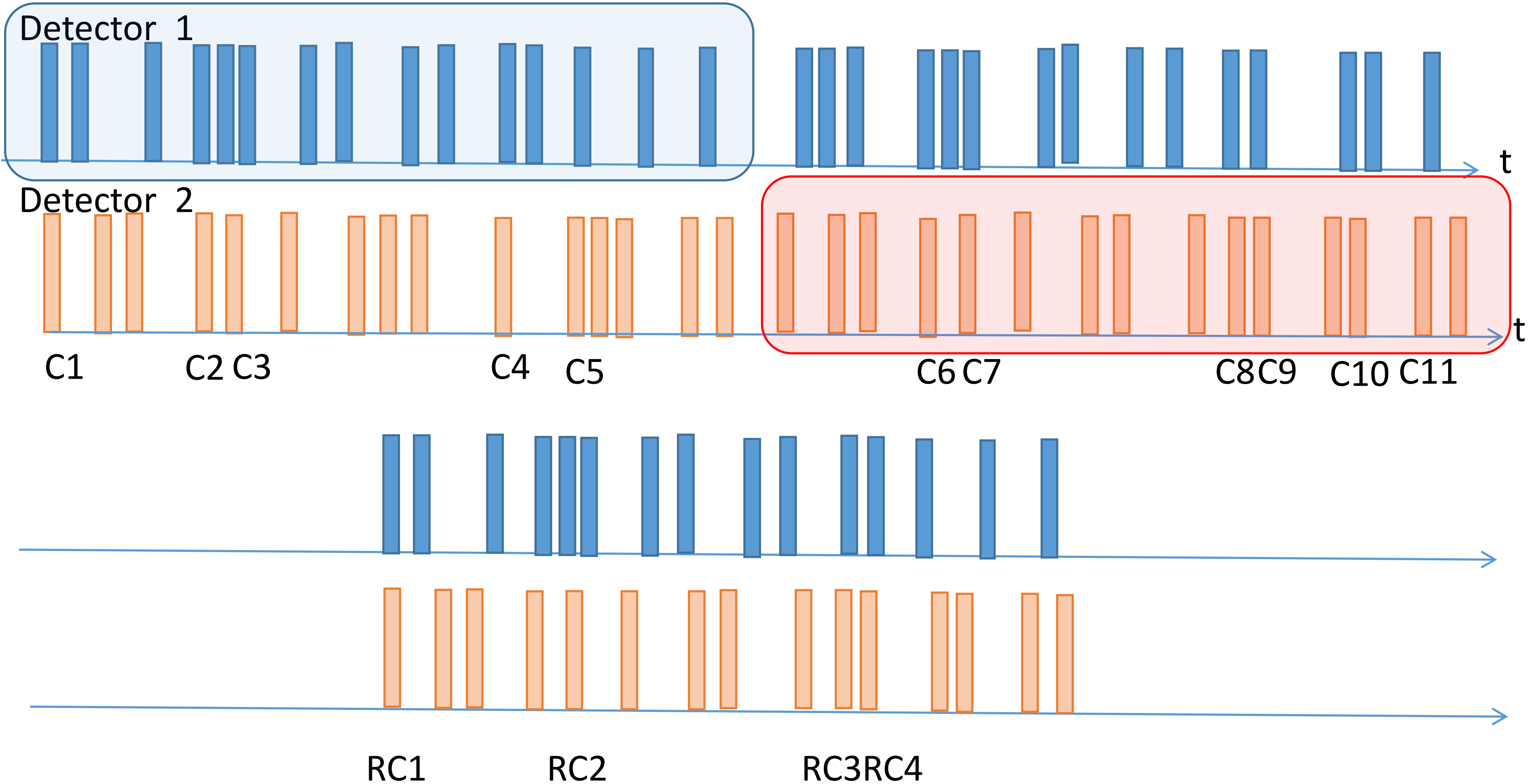
Delayed chain processing



- **A Delayed time window is set (R~50 ns)**
- **Artificial coincidences are generated with single events with events occurring at: $R \leq \Delta t \leq R+\tau$**

Delayed time window random coincidence estimation



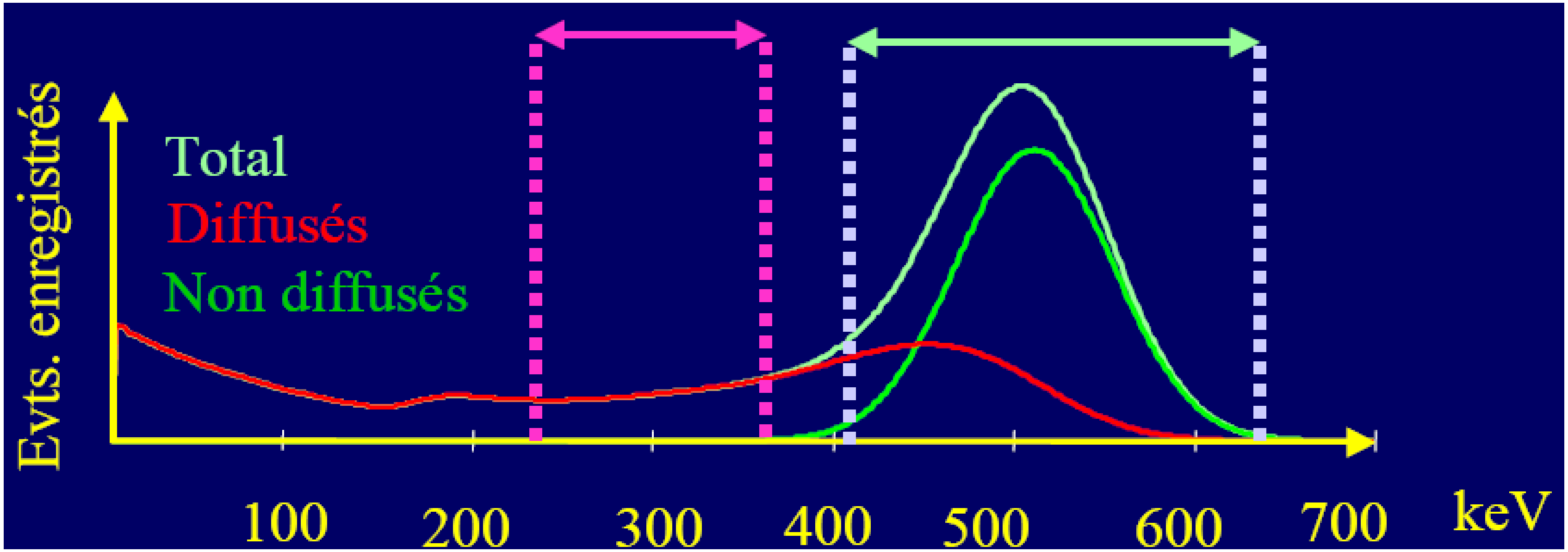


Random coincidence =8/11 = 73%

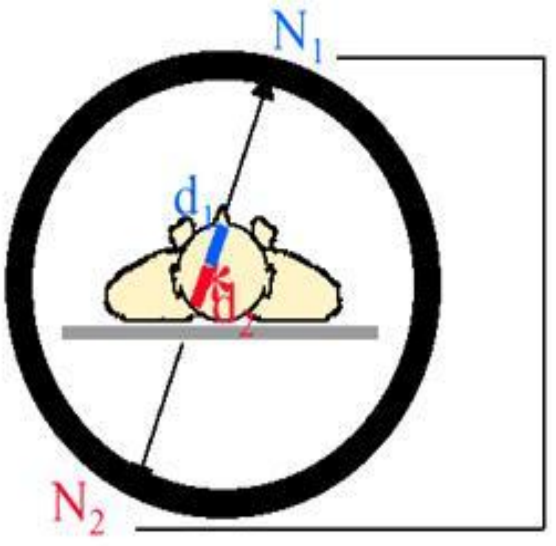
Scattered coincidence estimation

- By narrowing the coincidence energy window (not sufficient)
- Dual-energy windows (similar to SPECT) for scatter component estimation
- Analysis of the signal in LOR not intersecting the patient (CT/or transmission data available) to estimate the scatter fraction in LOR thought the patient
- Scatter contribution estimated by Monte Carlo (SSS = single scatter simulation)
- Possible to introduce the scatter correction into the system matrix in iterative reconstruction process

Scatter estimation with dual energy window



Attenuation correction



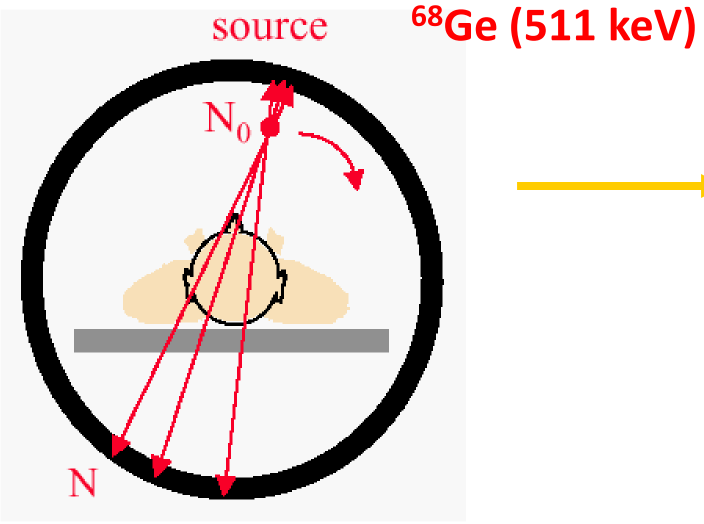
$$N_1 = N_\beta \exp\left(-\int_0^{d_1} \mu(L)\,dL\right)$$

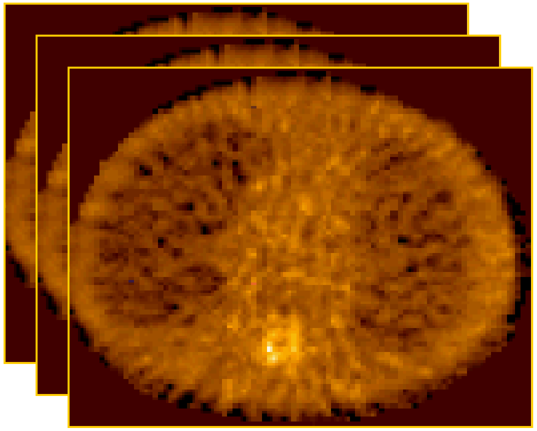
$$N_2 = N_\beta \exp\left(-\int_0^{d_2} \mu(L)\,dL\right)$$

$$N_1 N_2 = N_\beta^2 \exp\left(-\int_{d_1}^{d_2} \mu(L)\,dL\right)$$

- Unlike SPECT, the probability that both photons reach the detector is independent of the source location along the LOR
- Unlike SPECT, PET data can be accurately corrected for attenuation by simply multiplying each projection line by the appropriate ACF
- Attenuation corrections for PET require an attenuation map (μ-map, transmission image) at 511 keV

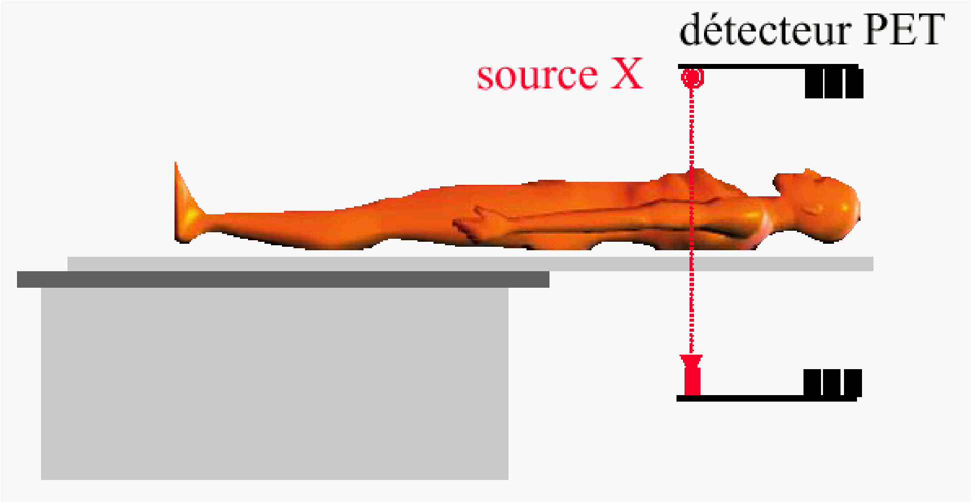
Historical: Attenuation Correction with transmission sources (no in use at present)

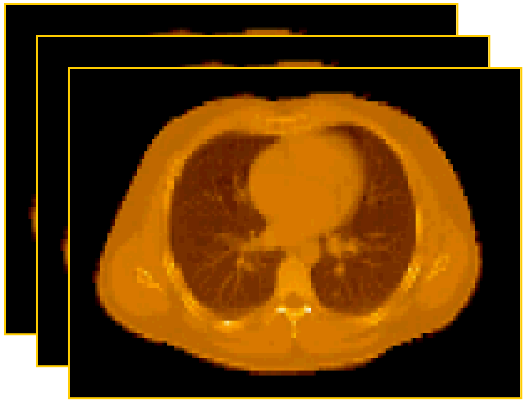




Transmission Scan

PET/CT → CT-derived attenuation map





Attenuation Correction

Transmission scan

The attenuation correction factor for a detector pair (i,j) is given by

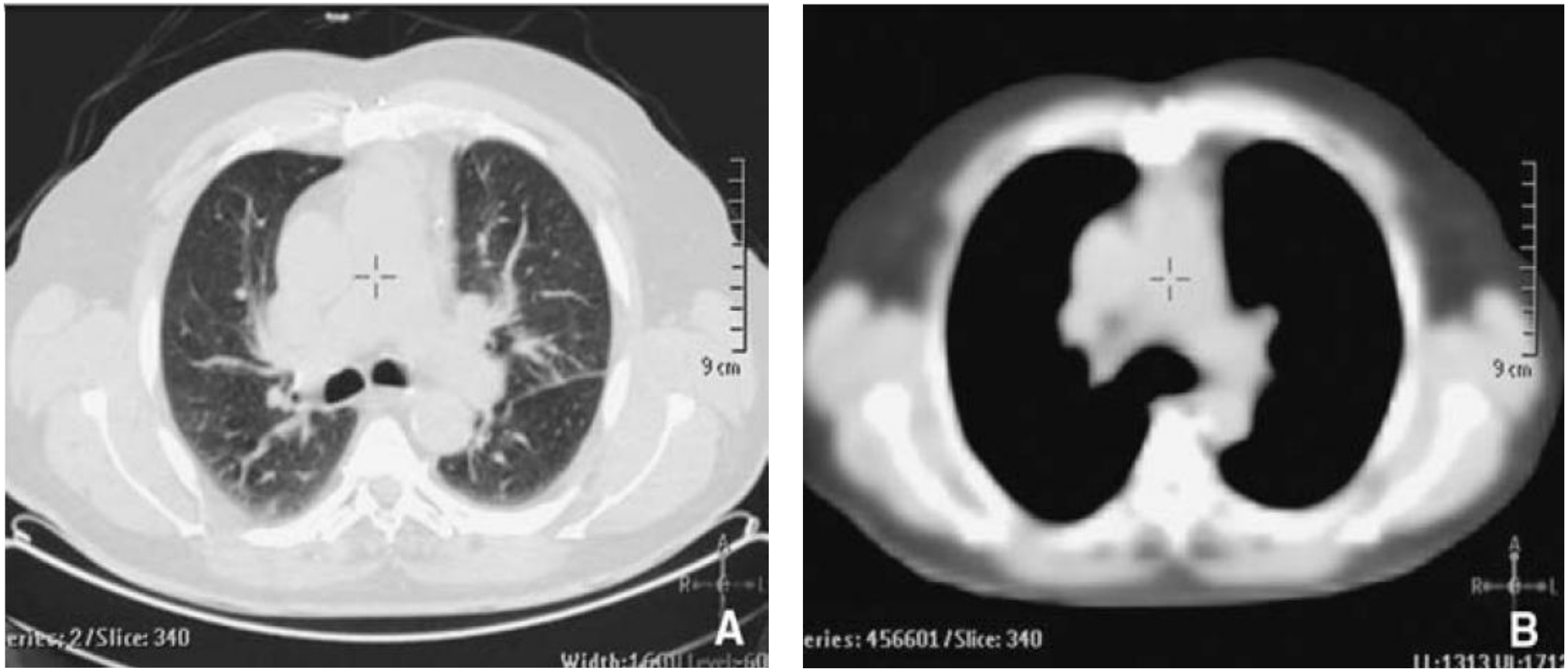
$$ACF_{ij} = e^{\mu L_{ij}} = \frac{Counts_{ij}^{BLANK}}{Counts_{ij}^{TRANSMISSION}}$$

CT scan

$\mu_{70keV} \rightarrow \mu_{511keV}$

1) Adjust resolution in CT (512x512, mm voxel size) to agree with resolution of PET (256 x 256 ~3 mm voxel size)
2) Convert CT numbers to $\mu_{70\ keV}$: $\mu = ((CT/1000)+1)*\mu_{H2O,70\ keV}$
3) Scale all μ values corresponding to CT values below ≈ 200 − 300 by a factor of $\mu_{H2O,511\ keV}/\mu_{H2O,70\ keV}$
4) Scale all μ values corresponding to CT values above ≈ 200 − 300 by a factor of $\mu_{bone,511\ keV}/\mu_{bone,70\ keV}$

Conversion from CT-number to μ-map in PET

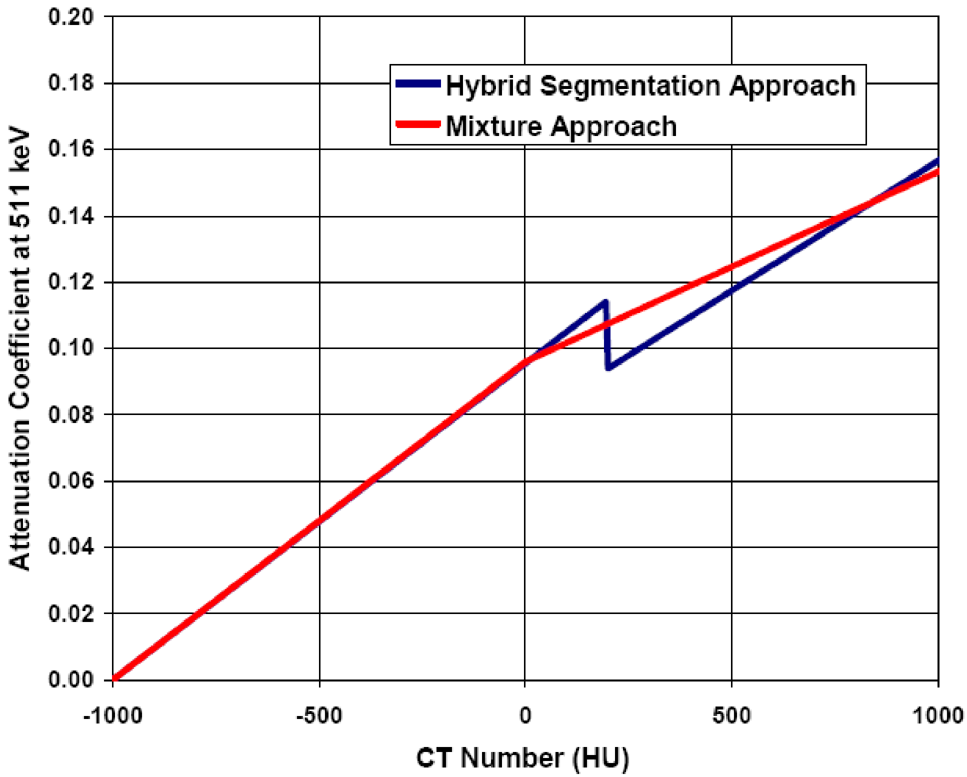


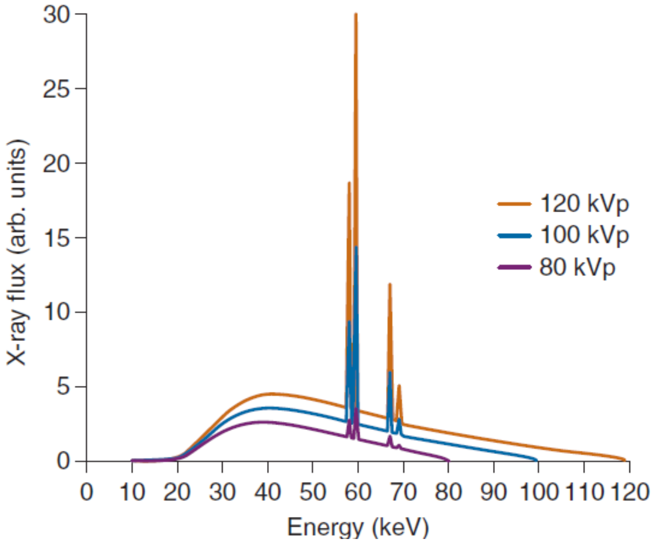
Noise in transmission data → Induced noise on attenuation correction data

Segmentation: assumption of homogeneous μ into the segmented regions, typically soft tissue, bones and lungs

Image smoothing

Conversion from CT-number to μ-map in PET

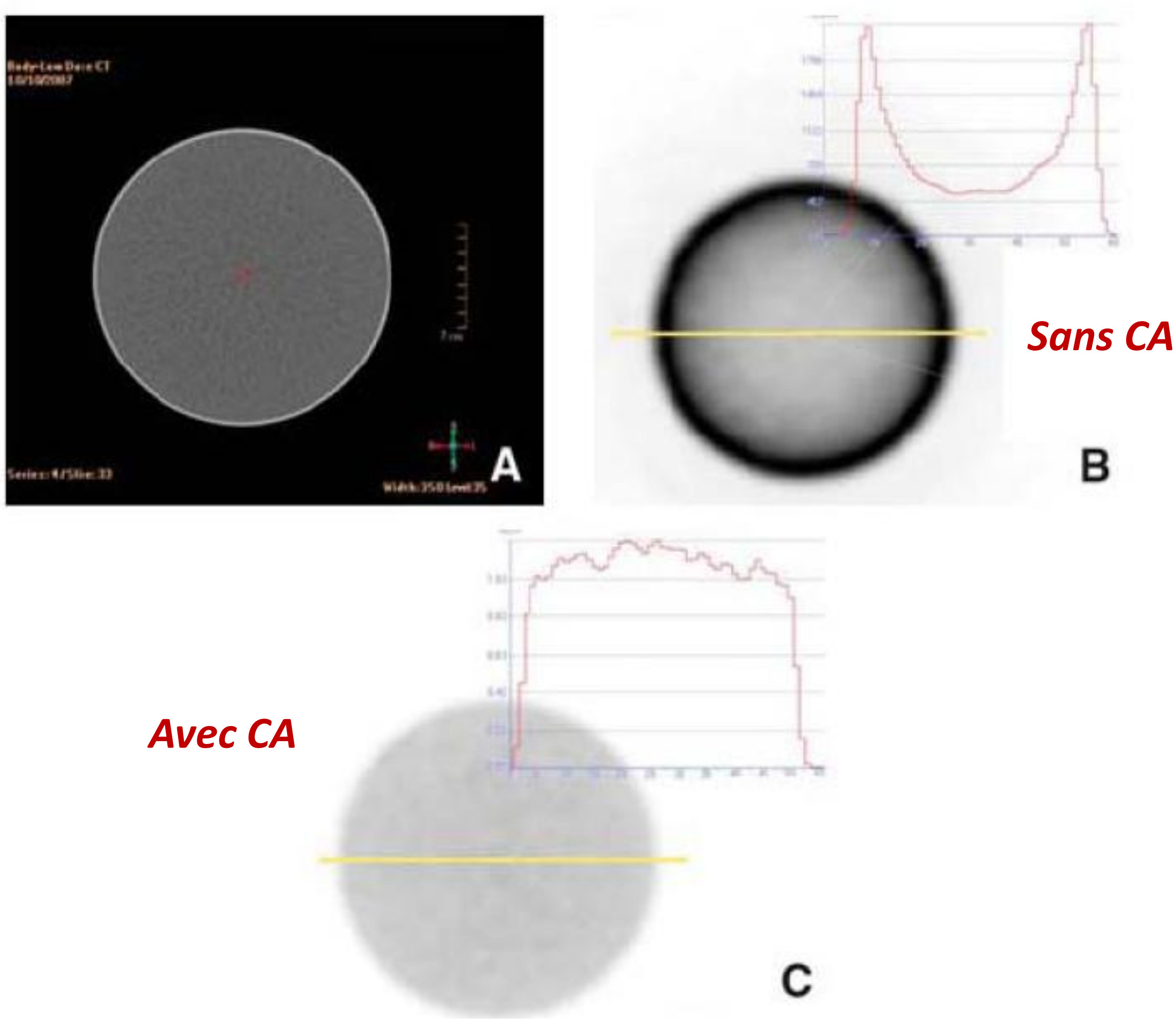




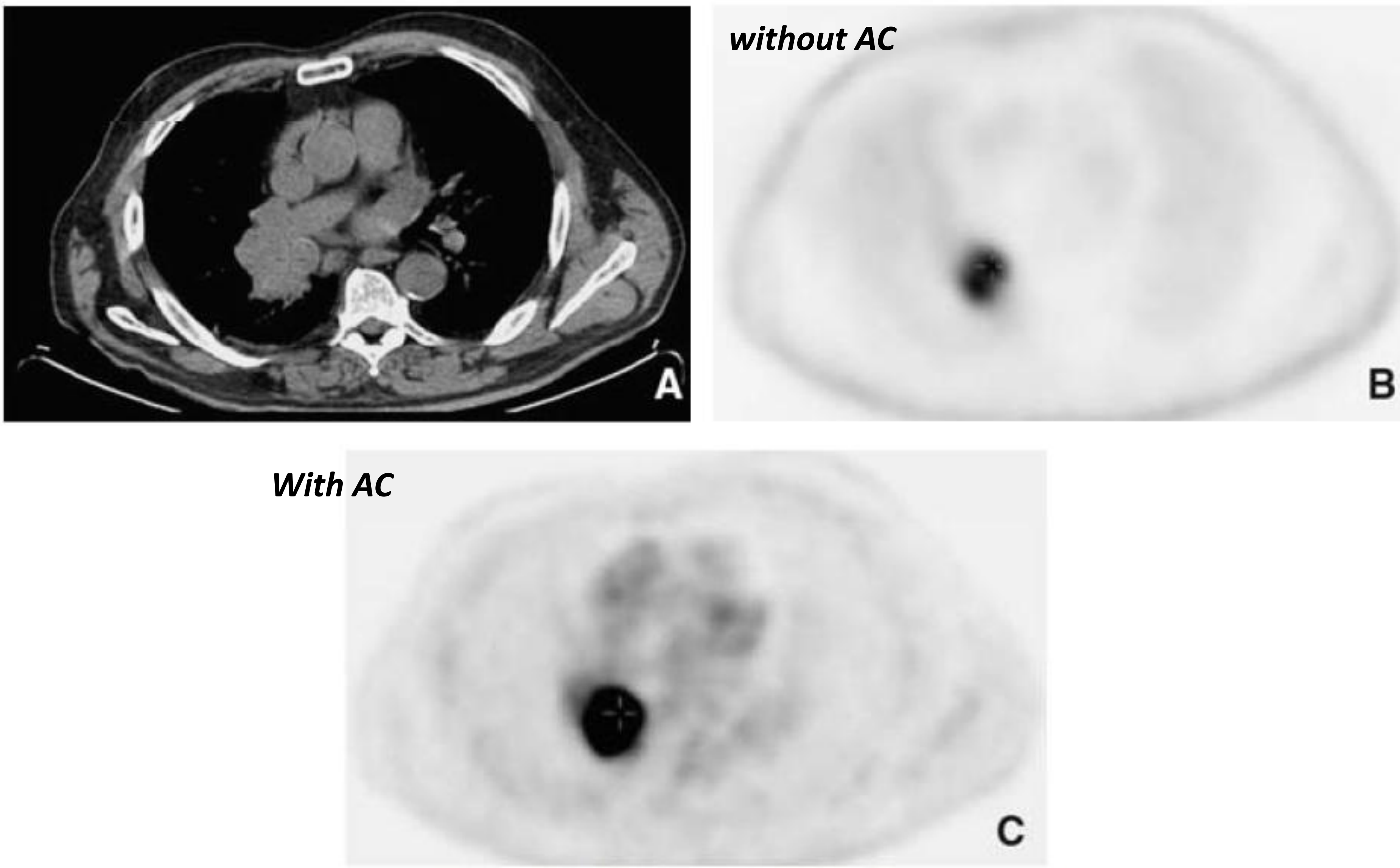
Mixture

- Threshold = 0 HU
- $\mu_{H2O,80\ keV} = 0.184\ cm^{-1}$
- $\mu_{bone,80\ keV} = 0.428\ cm^{-1}$
- $\mu_{H2O,511keV} = 0.096\ cm^{-1}$
- $\mu_{bone,511\ keV} = 0.172\ cm^{-1}$

Attenuation correction
in a simple cylindrical homogeneous configuration



Attenuation correction in a patient case



Corrections for Quantitative Studies : All PET is (almost) Quantitative !

Raw Sinogram Data (Trues + Scatters + Randoms)

Remove Randoms

Normalize Detector Responses

Correct for Deadtime

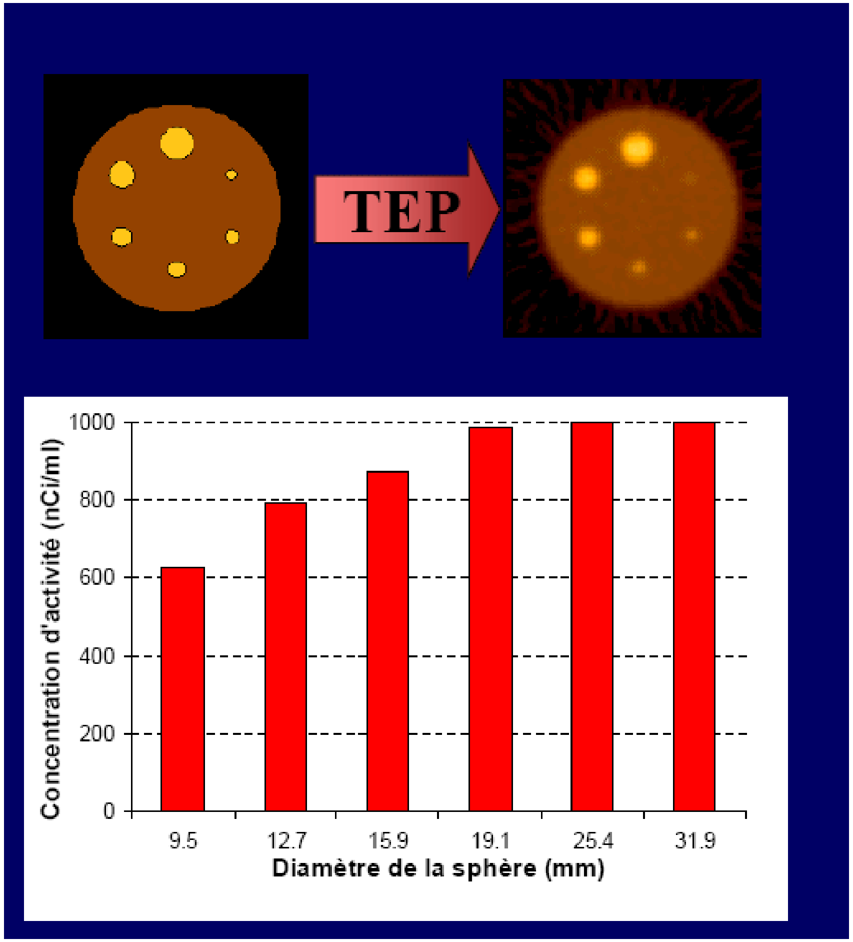
Correct for Scatter

Correct for Attenuation

Sinogram
Ready for Reconstruction

It's not exactly like this, and it's not necessarily as linear as this!

Partial volume effect (PVE)



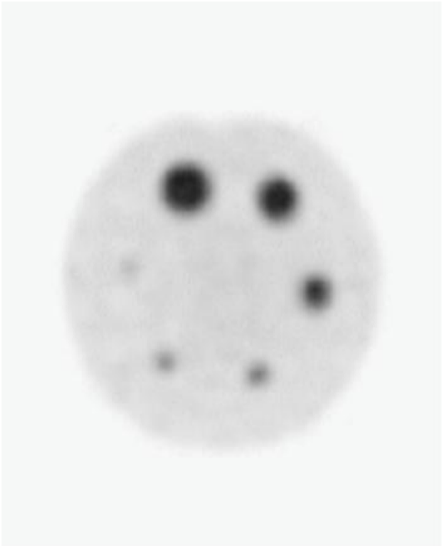
Activity concentration underestimation occur when the lesion size is ~2-3 times the system spatial resolution.

PVE is stronger as the lesion size is smaller.

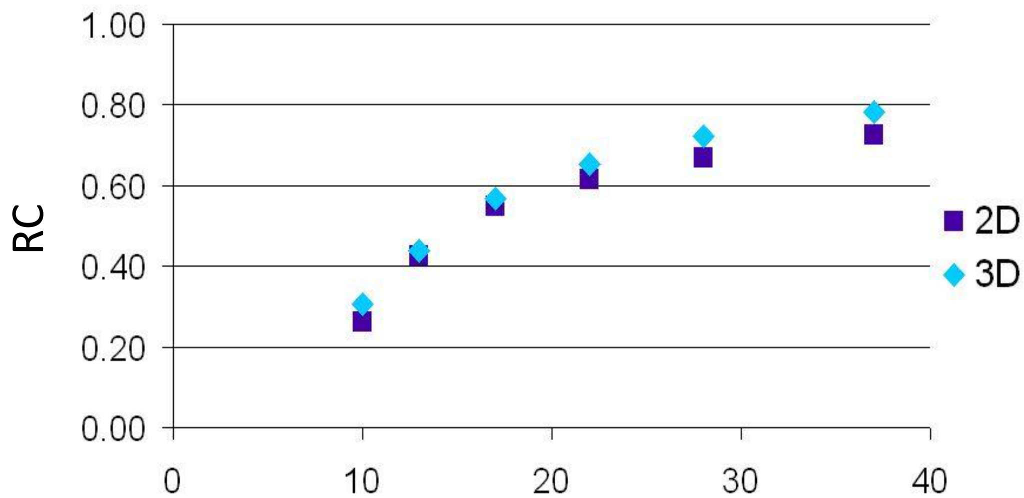
PVE correction

Recovery coefficient (RC)

$$RC = \frac{\text{Observed activity concentration}}{\text{True activity concentration}} \Rightarrow A_{corrected} = \frac{A_{measured}}{RC}$$



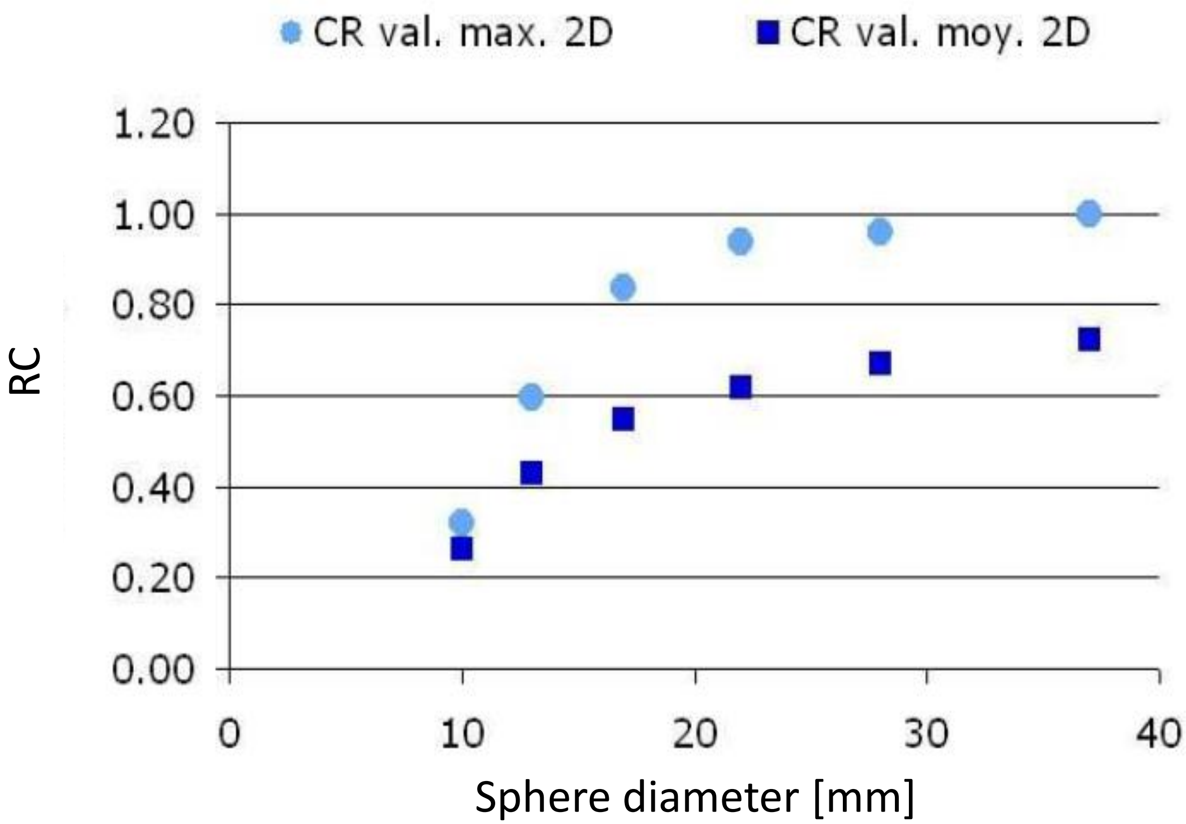
[18]*F (GE Discovery LS)*

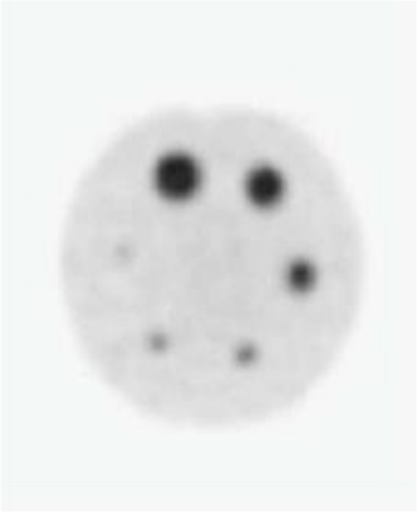


Sphere diameter [mm]

PVE correction

Two RC estimators: Rc max Rc_{ave} on the sphere volume





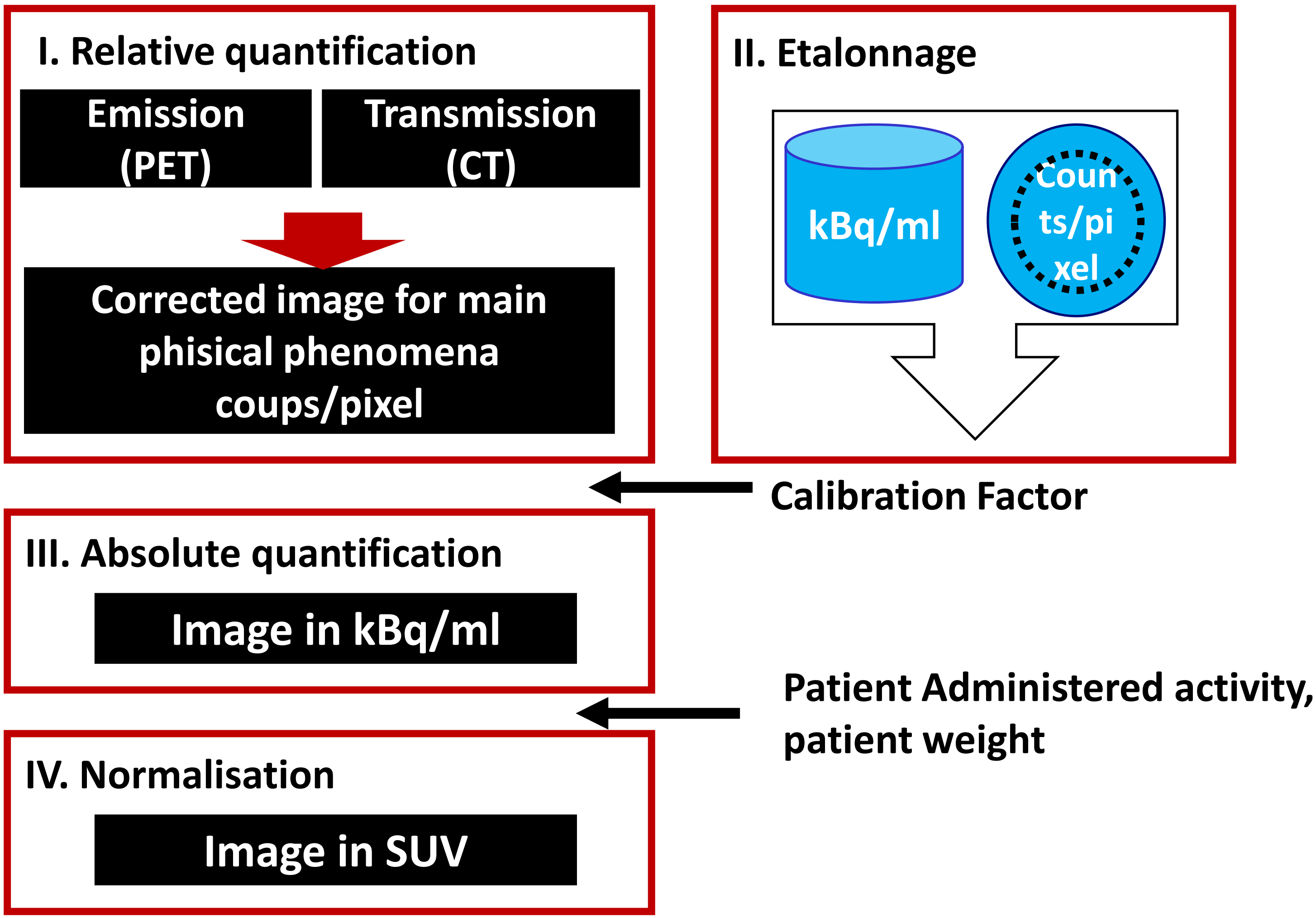
Absolute quantitation

- ≠ Relative quantification
  - Signal in a certain region compared to others
    - ex. tumor to healthy tissue ratios
- Aim: to have a quantitative estimate of the activity concentration (in a given region: organ, tissue) expressed in kBq/ml (with no reference to other regions)
- Goals :
  - Comparison between patients
  - Perform 3D (organ/tissue based or voxel based) dosimetry
  - Therapy follow-up; (response to therapy)
  - Dose/response ralation in radiotherapy
- Calibration factor :

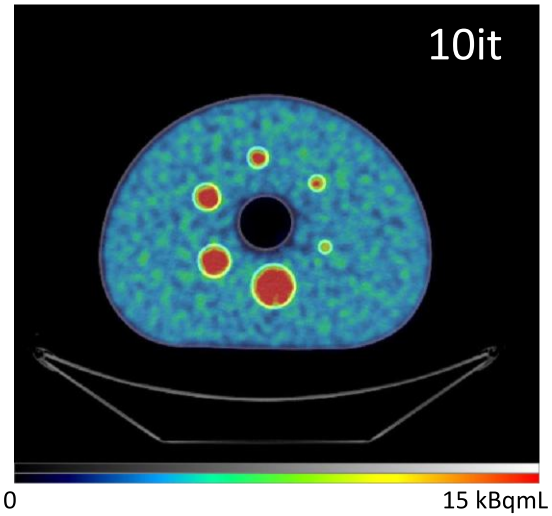
$$\textbf{CF in} \quad \frac{\textbf{Bq/ml}}{\textbf{cps/voxel}}$$

Ex. Using simple phantom configurations

Quantitative calibration workflow

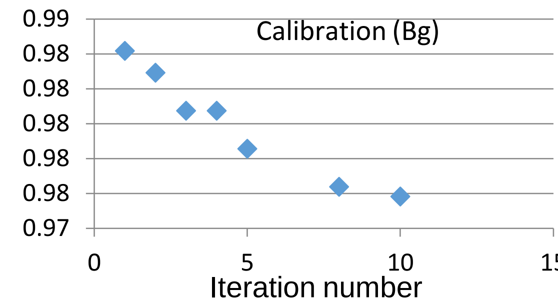


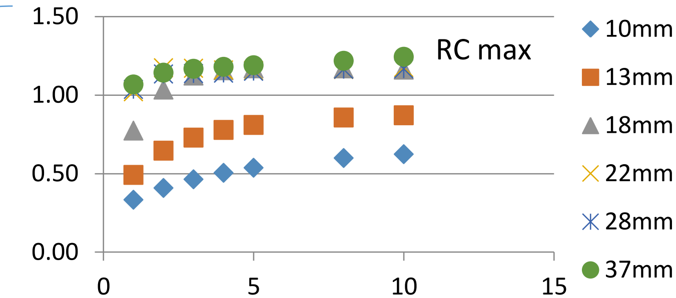
Example of Ga-68 PET optimization

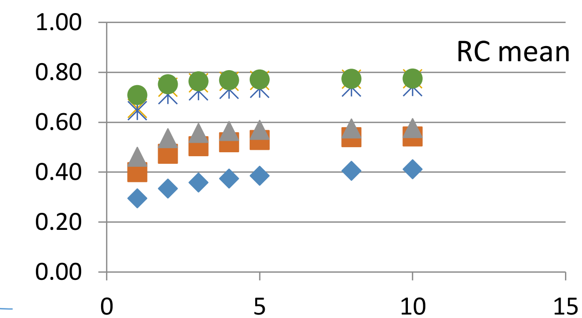


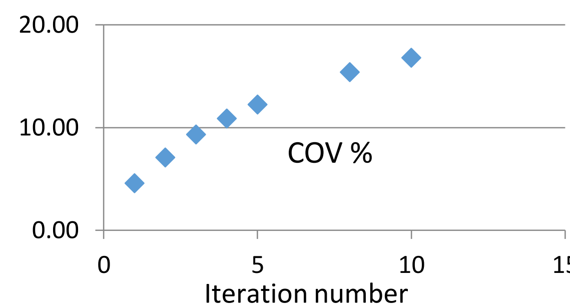
Signal recovery as a function of the iteration number

- Quantitative calibration (Bg)
- Recover coefficient (max, mean)
- Image noise (COV)

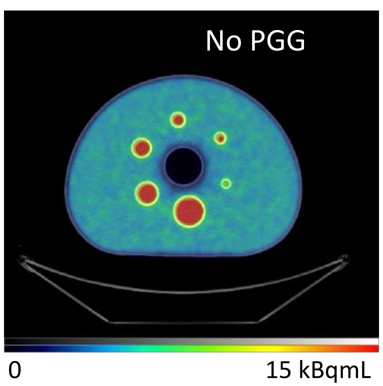


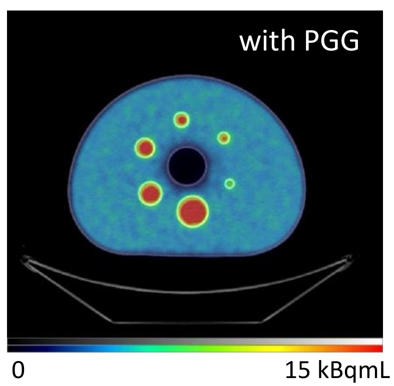






Quantitative assessment of prompt gamma correction (PGC)





If no PGC → 8% overestimation of the expected activity concentration

Specific Uptake Value (SUV) concept

$$\text{SUV} = \frac{\text{Measured Activity Concentration } [\text{Bq/ml}] \times 10^3}{\left[\dfrac{\text{Injected Activity* } [\text{Bq}]}{\text{Patient Mass } [\text{kg}]}\right]}$$

** Reported at the beginning of the exam*

SUV = semi-quantitative index of the ^{18}FDG accumulation ⇒ makes possible the comparisons between exams

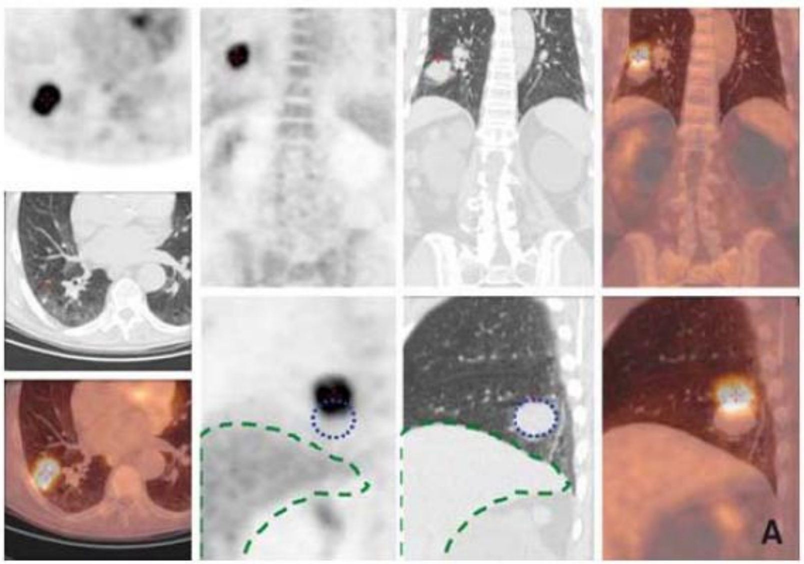
- SUV is used as an index to characterizhe region of specific uptake
- Its use depends on :

Time between injection and acquisition, patient's blood sugar level, patients's weight, quantification quality (attenuation correction, …), partial volume effects, …

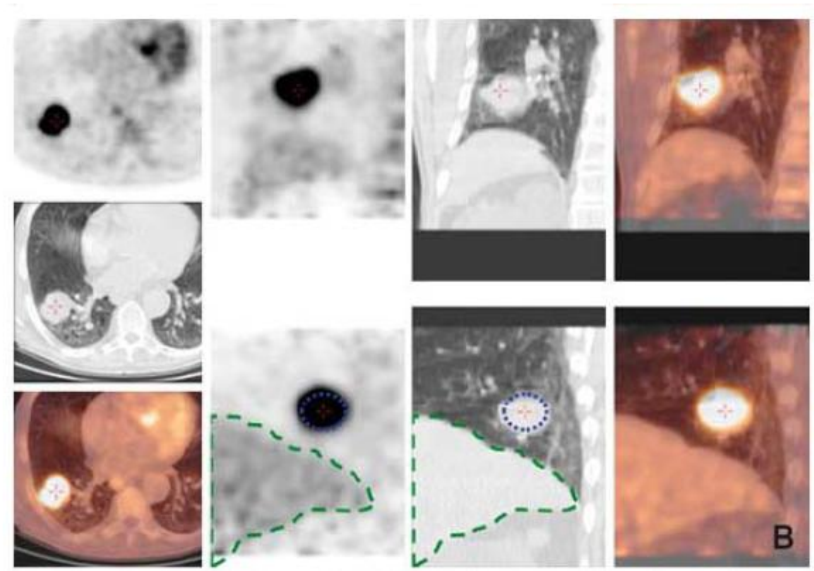
Confounding factors in PET/CT

- Patient motion
- Respiratory motion
  - PET : normal breath (minutes)
  - CT : breath hold can be used (~sec)
- Use of contrast agents
- Partial FOV overlap (CT FOV < PET FOV)
- Low space resolution (PVE)

PET with synchronized respiratory gating

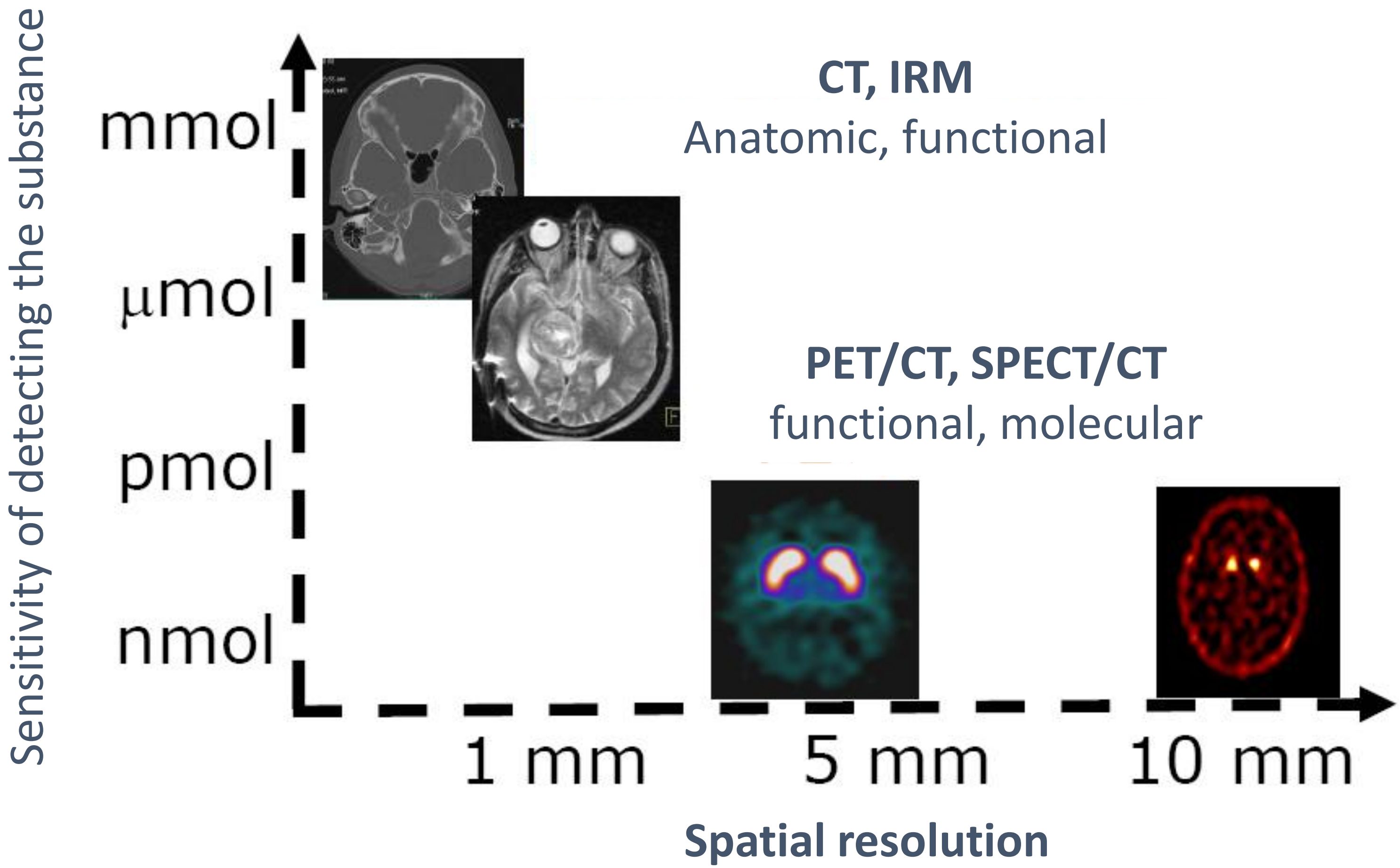


Without respiratory syncronisaton



With respiratory syncronisaton

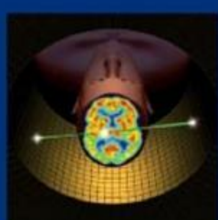
Short comparison of imaging modalities



Physics in Nuclear Medicine

FOURTH EDITION

Simon R. Cherry
James A. Sorenson
Michael E. Phelps



SAUNDERS
ELSEVIER