

Imagerie en Médecine Nucléaire

Irène Buvat

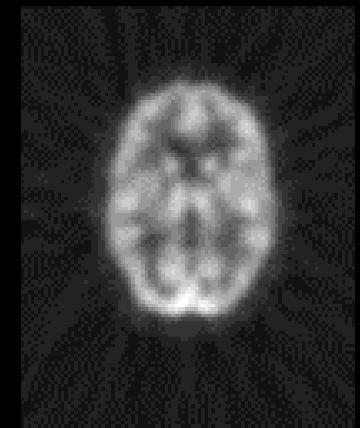
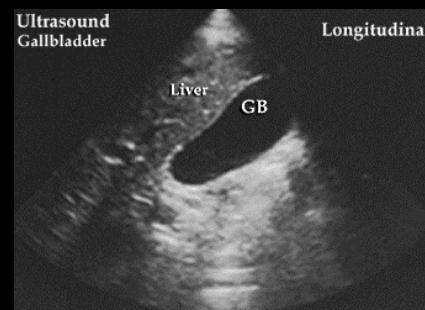
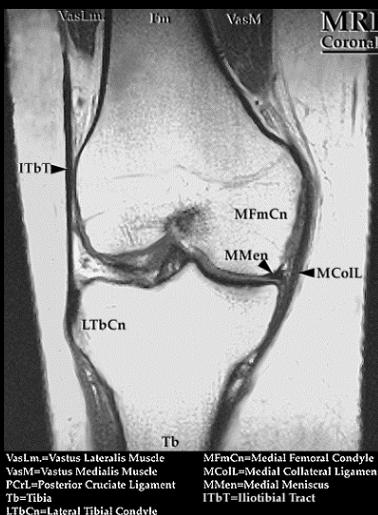
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<http://www.guillemet.org/irene>

Introduction

- 1895

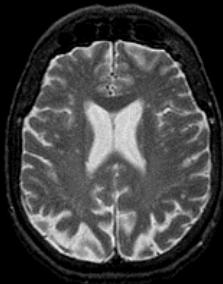


- Aujourd'hui



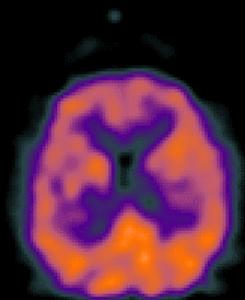
Deux types d'imagerie

- Imagerie morphologique : étude de l'anatomie



scanner, IRM, ... :
haute résolution spatiale

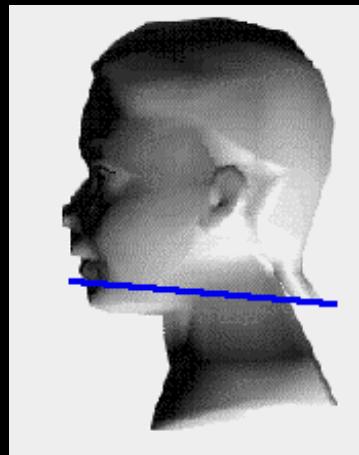
- Imagerie fonctionnelle : étude de processus biochimiques et physiologiques



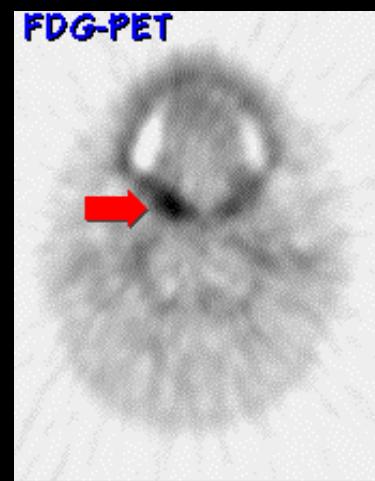
imagerie scintigraphique :
résolution spatiale médiocre

Intérêt de l'imagerie fonctionnelle

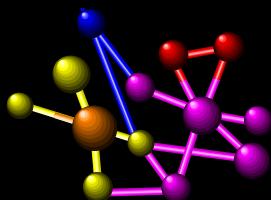
- Imagerie anatomique normale



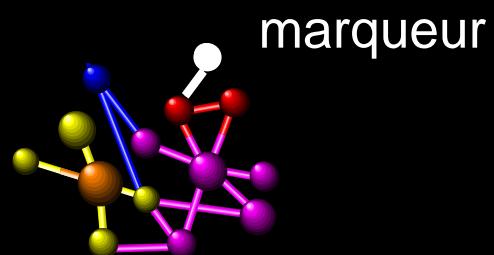
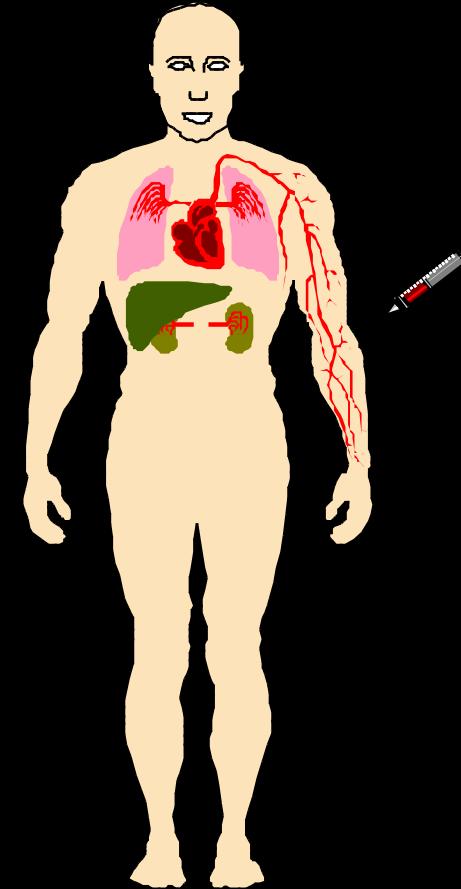
- Imagerie fonctionnelle anormale



Principe général de l'imagerie fonctionnelle



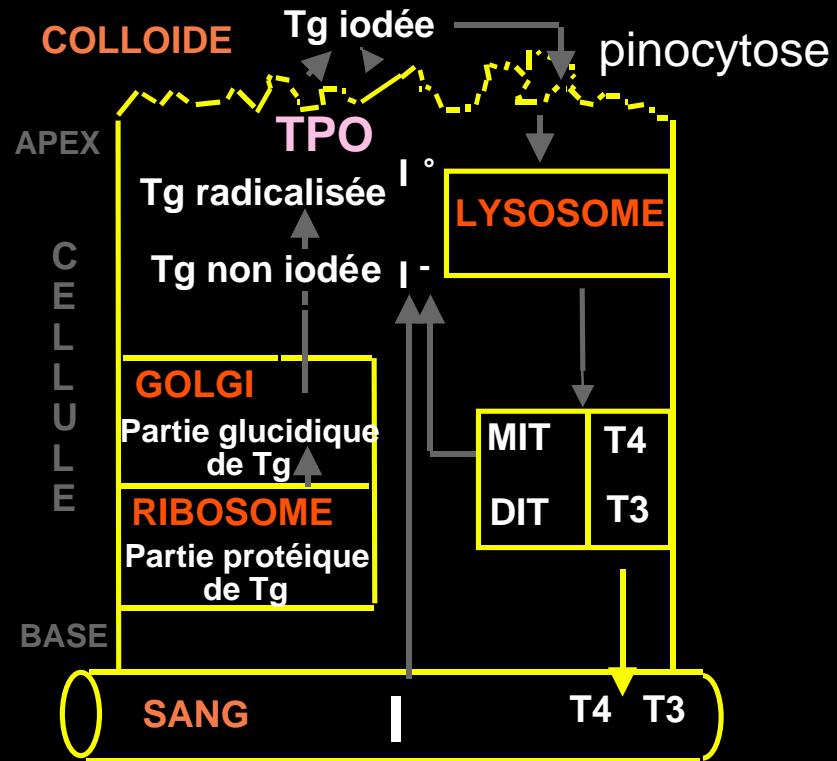
1. Traceur caractéristique d'une fonction métabolique ou physiologique



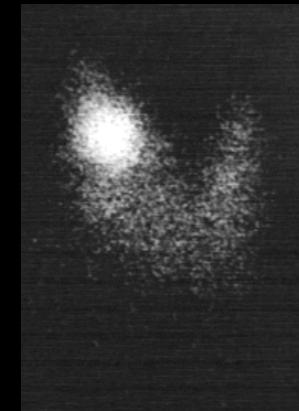
2. Marquage

3. Etude du devenir de la molécule marquée

Principe de l'imagerie fonctionnelle



2. Isotope : I-123

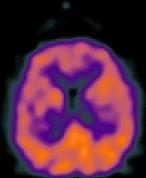
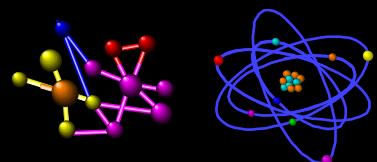


1. Matière première indispensable à la fabrication des hormones thyroïdiennes : Iode

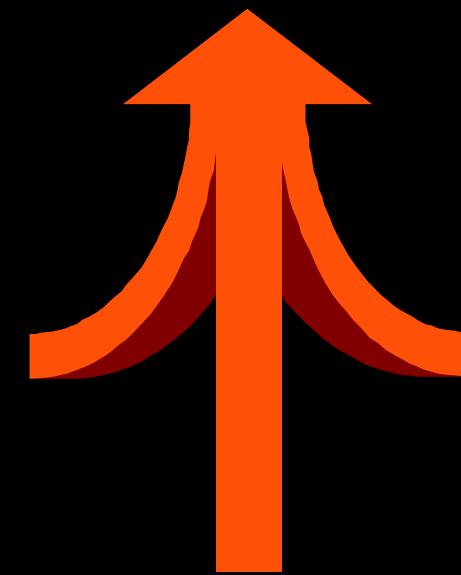
3. Etude du devenir de la molécule

Les pierres d'angle

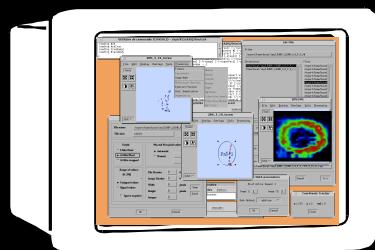
Radiotraceur



Détecteur



Traitement de l'information



Radiotraceurs



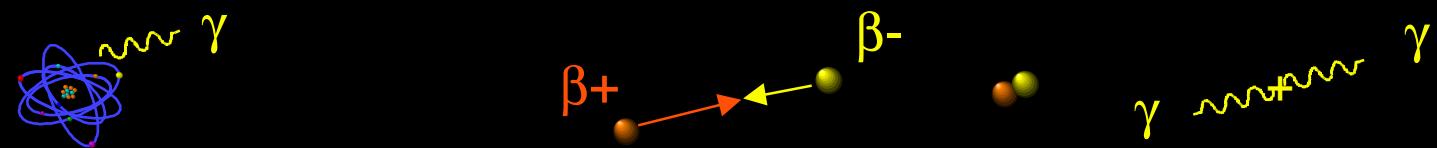
- Propriétés chimiques :
 - conservation des propriétés de la molécule considérée
 - concentration nanomolaire
 - liaison forte
- Exemples de traceurs : molécules, cellules, ligands, hormones, etc

Radiotraceurs



- Propriétés physiques :

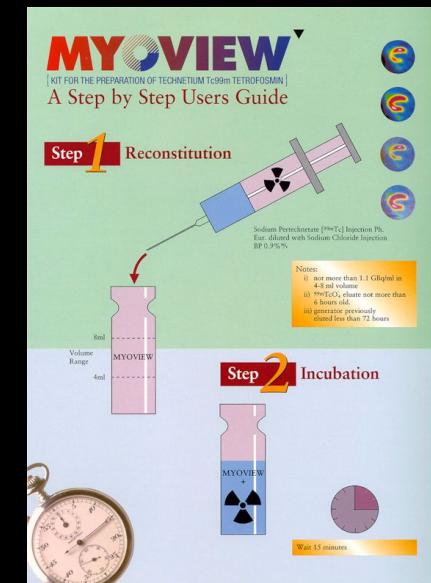
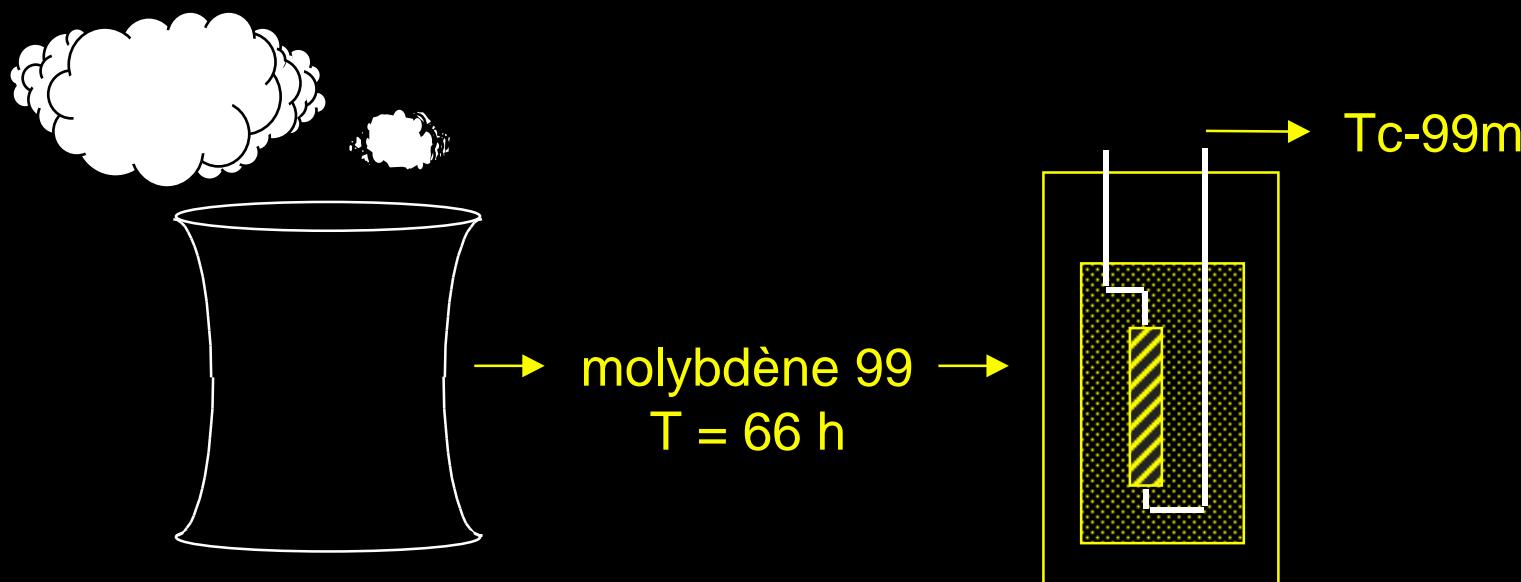
- nature du rayonnement émis :



- énergie du rayonnement émis : 50 - 511 keV
 - période du radioisotope : qq minutes à qq dizaines d'heures

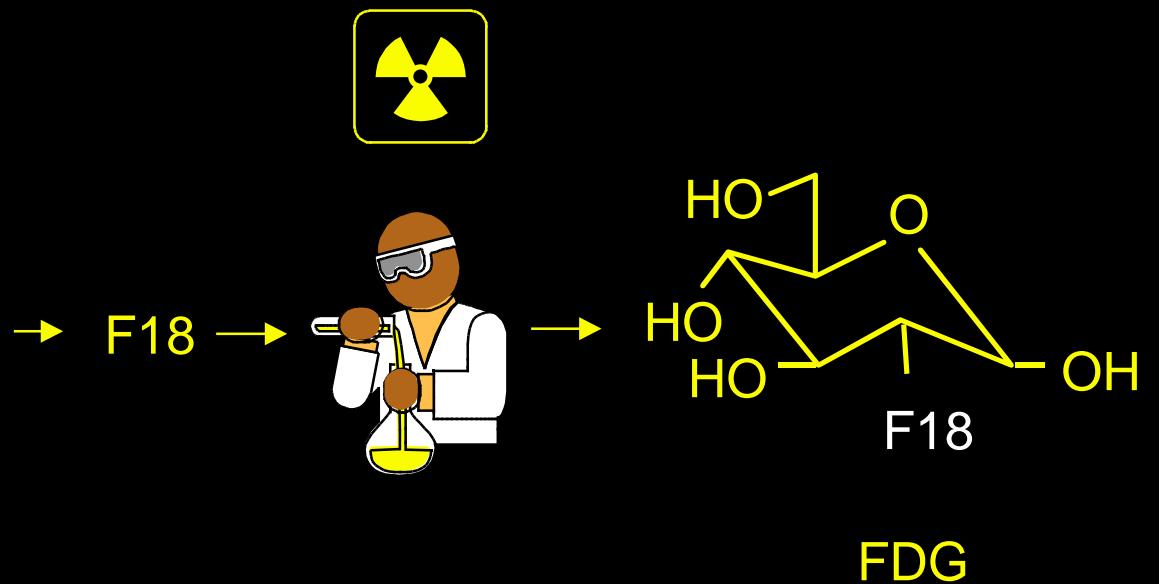
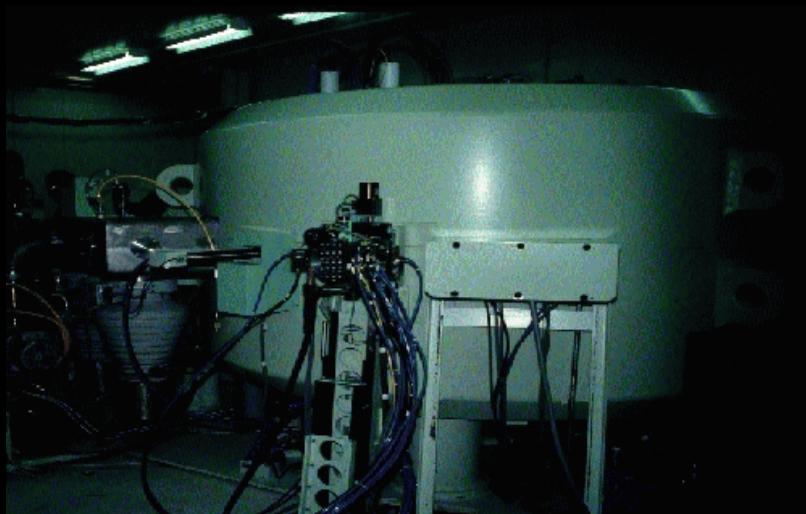
Isotope émetteur de photons γ le plus utilisé : le Tc-99m

Technétium 99m : énergie = 140 keV $T = 6 \text{ h}$



Isotope émetteur de positons le plus utilisé : le F18

Fluor 18 : $T = 110 \text{ min}$

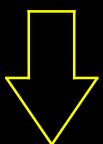
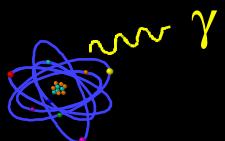


Isotopes utilisés en imagerie nucléaire

Isotope	Energie	Période
Emetteurs de photons γ		
Technétium 99m	140 keV	6 heures
Iode 123	159 keV	13 heures
Thallium 201	71 keV	73 heures
Emetteurs de positons β^+		
Oxygène 15	511 keV	2 minutes
Azote 13	511 keV	10 minutes
Carbone 11	511 keV	20 minutes
Fluor 18	511 keV	110 minutes
Brome 76	511 keV	978 minutes

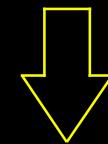
Types d'émetteurs et techniques d'imagerie

Emetteurs de γ



- scintigraphie planaire
- tomographie d'émission monophotonique : SPECT

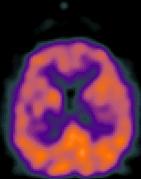
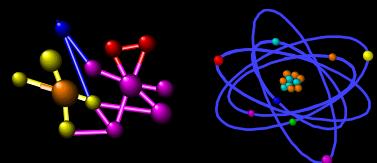
Emetteurs de positons



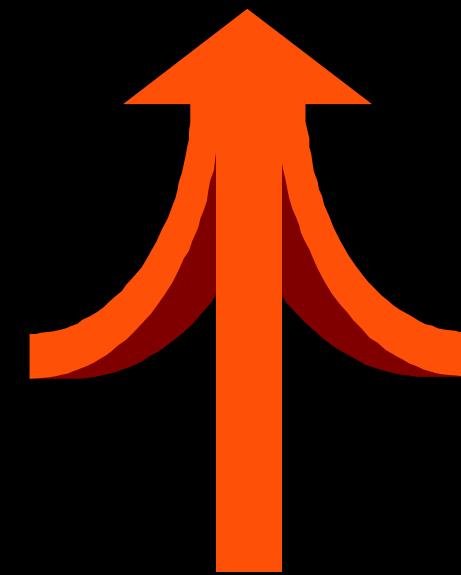
- tomographie par émission de positons : PET

Les détecteurs

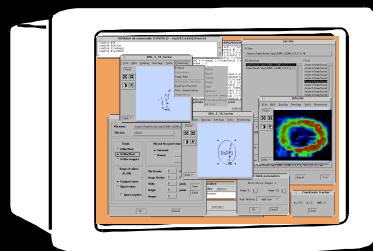
Radiotraceur



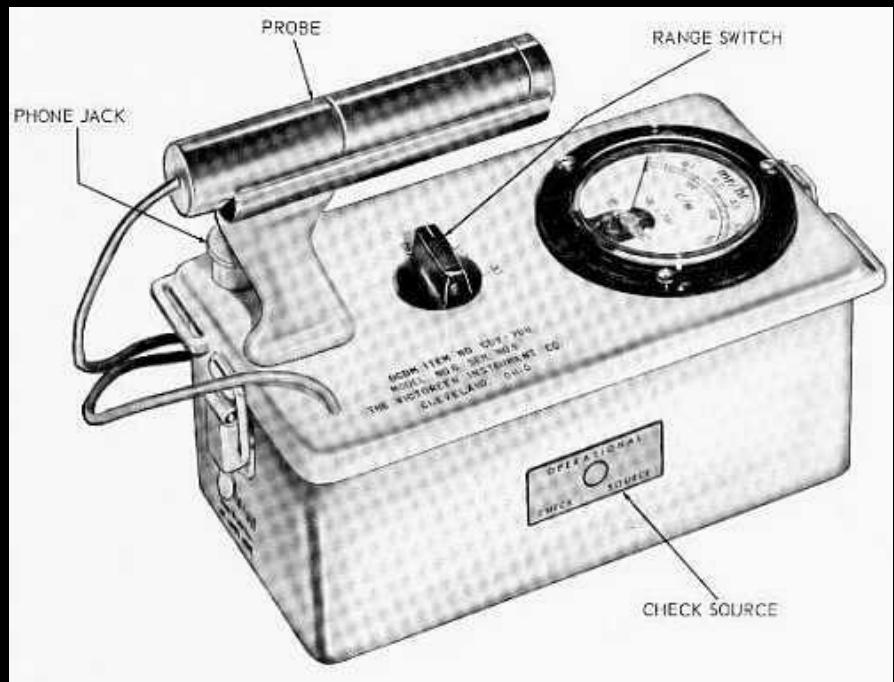
Détecteur



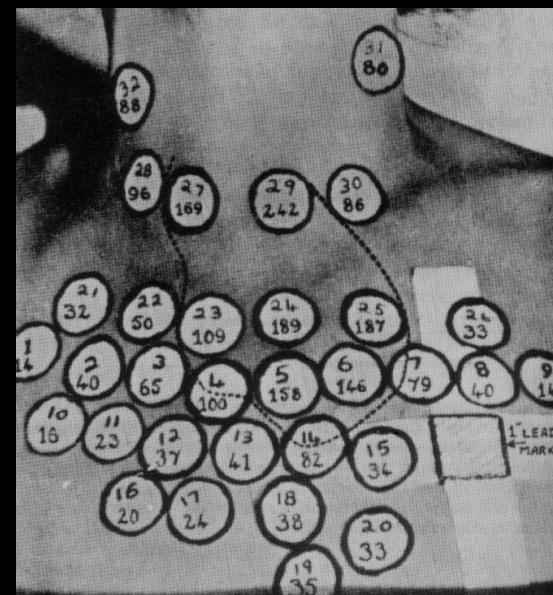
Traitement de l'information



1948 : Comptage des rayonnements gamma point par point

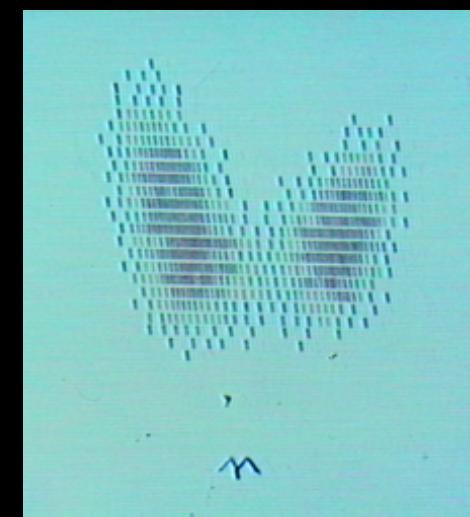
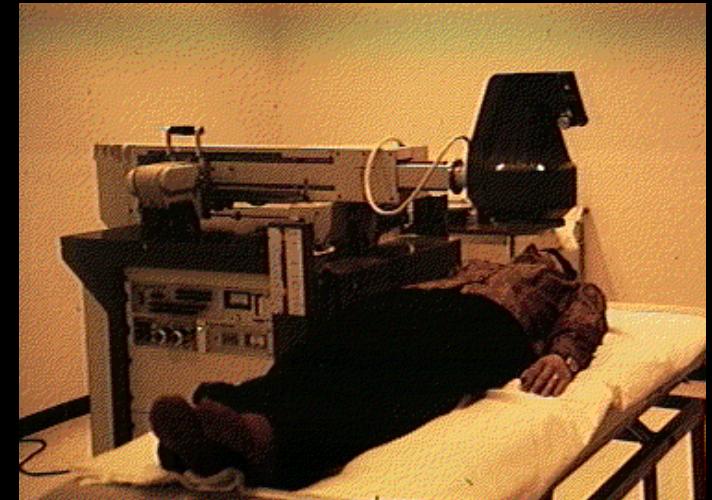
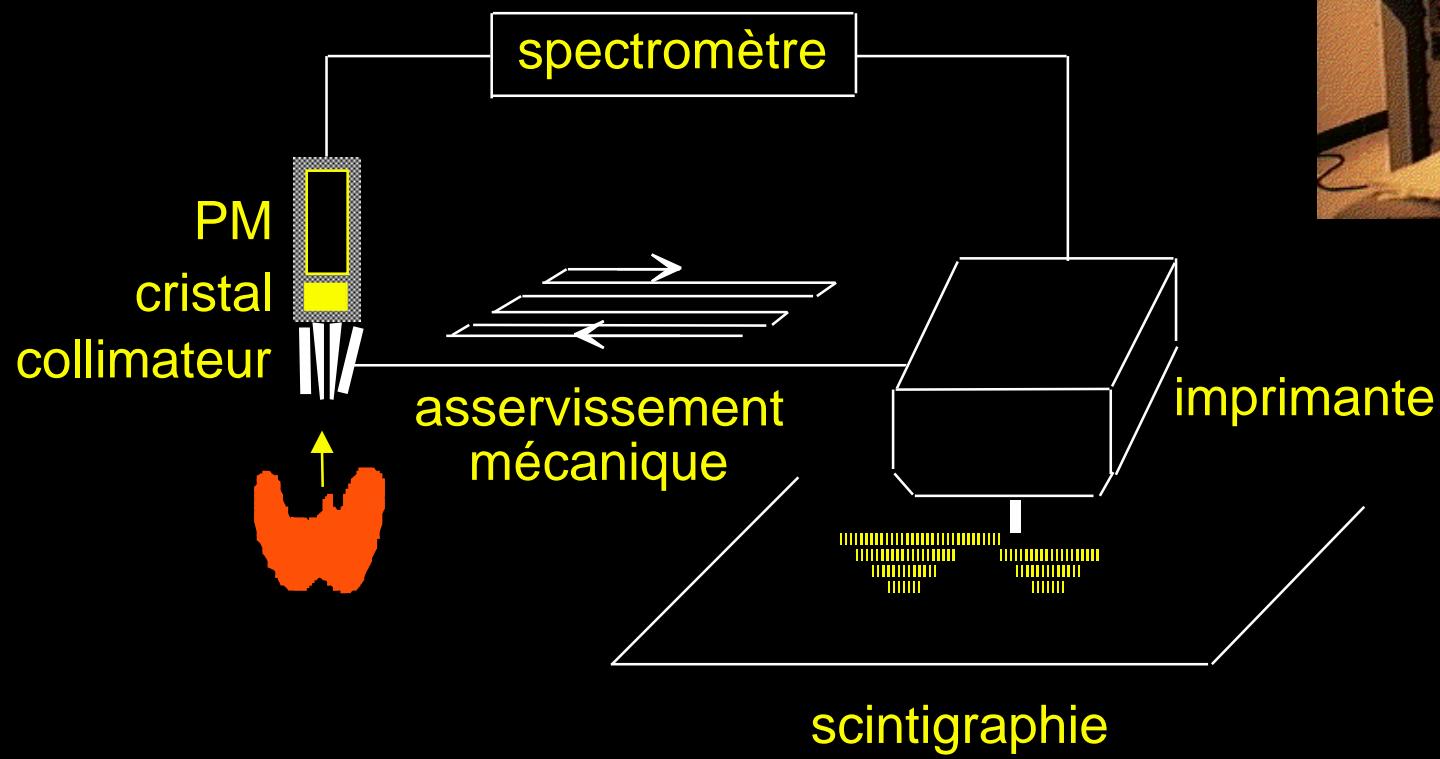


Compteur Geiger-Müller

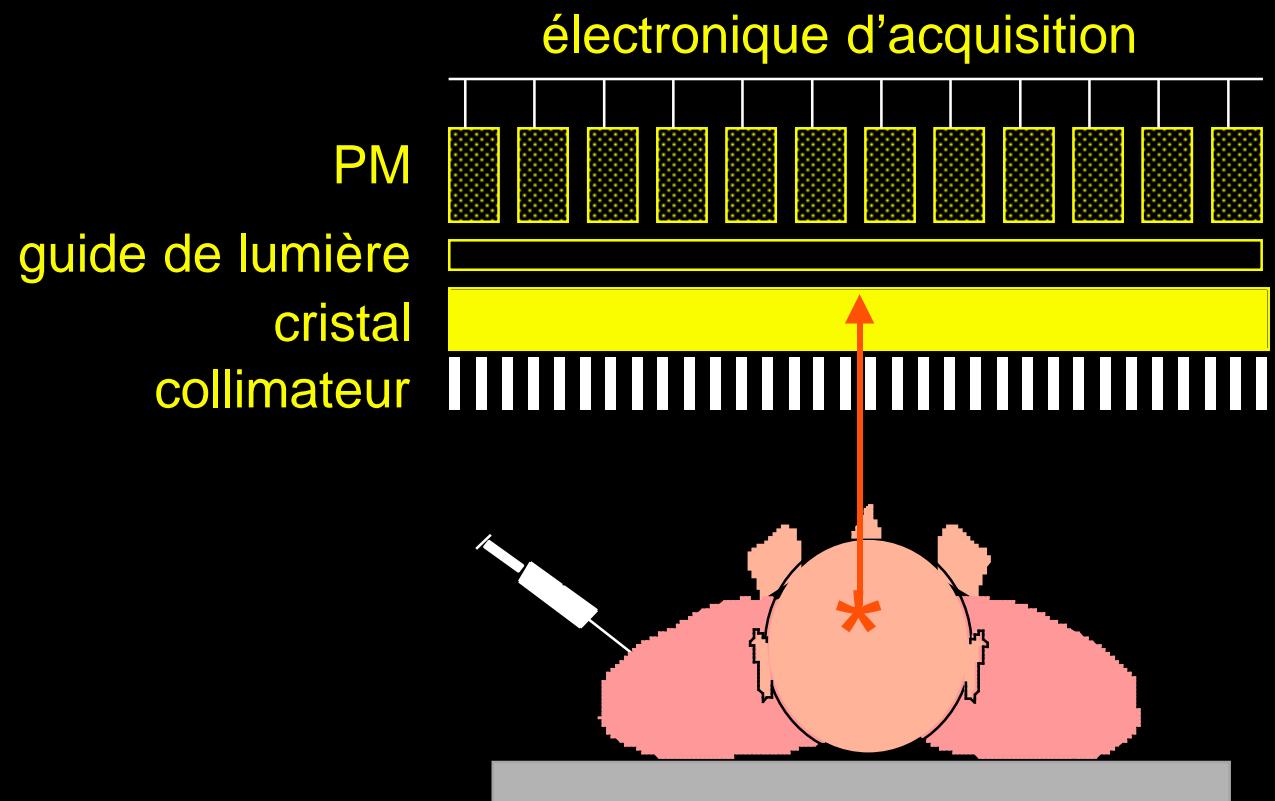
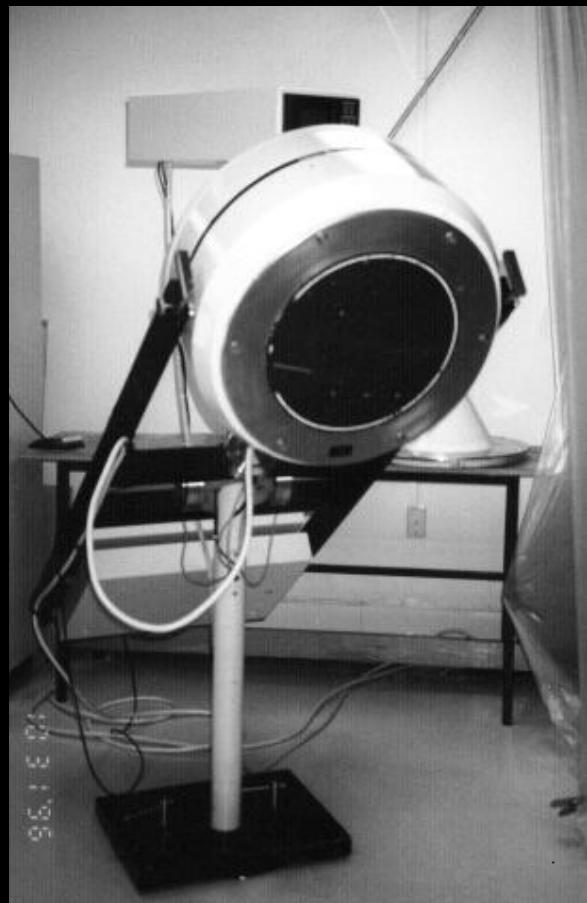


I-131

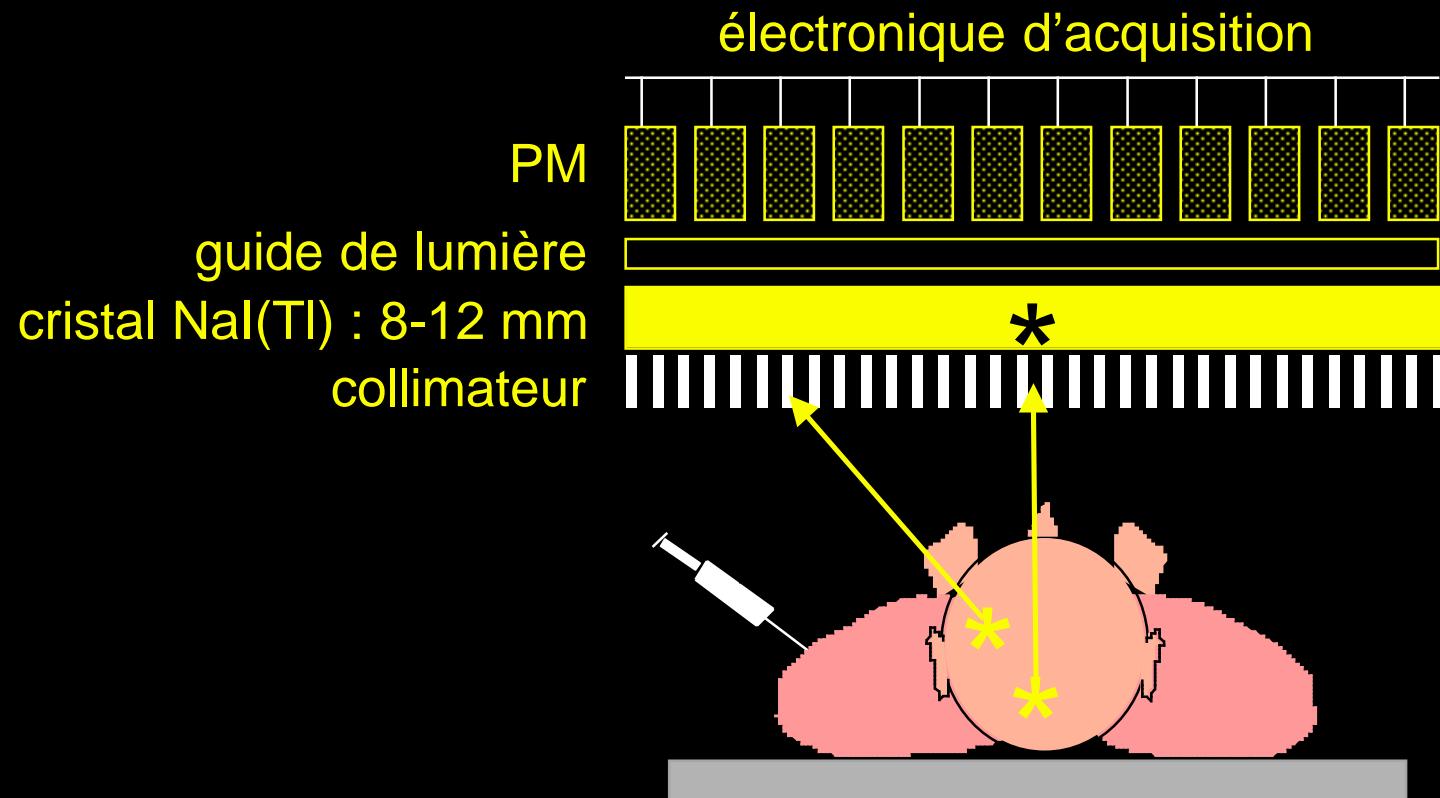
1951 : Scintigraphe à balayage



1958 : Gamma caméra



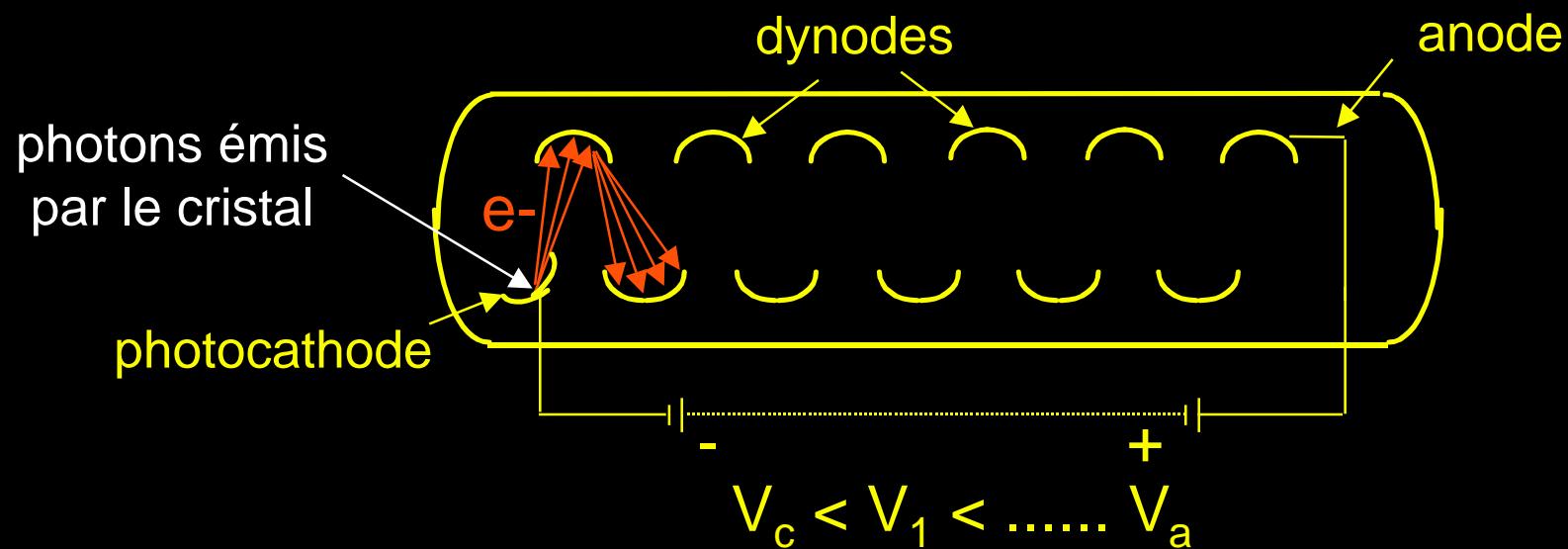
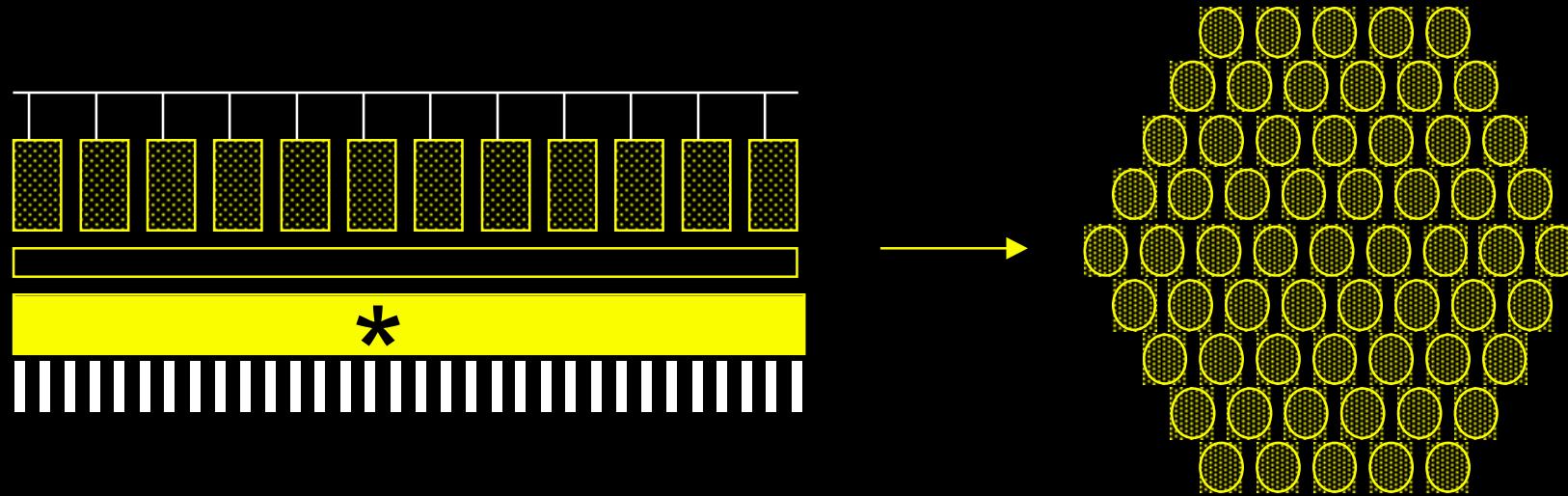
Principe de la gamma caméra



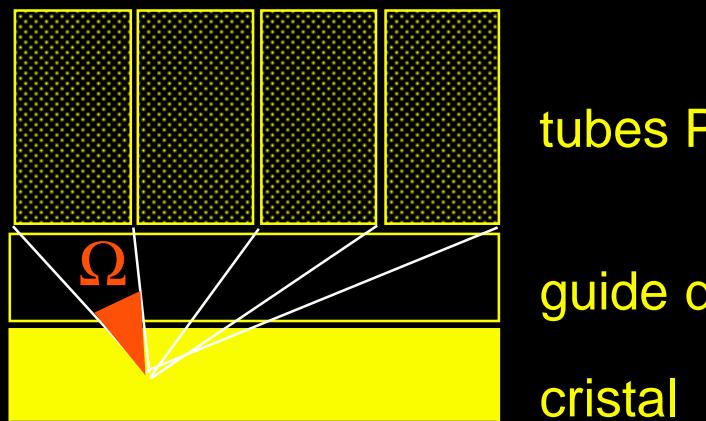
NaI(Tl) :

~ 430 nm ~3 eV = lumière bleue-verte
densité : 3,7 g/cm³
constante de décroissance : 230 ns => 2000 cps / PM
rendement lumineux : 13%

Tubes photomultiplicateurs



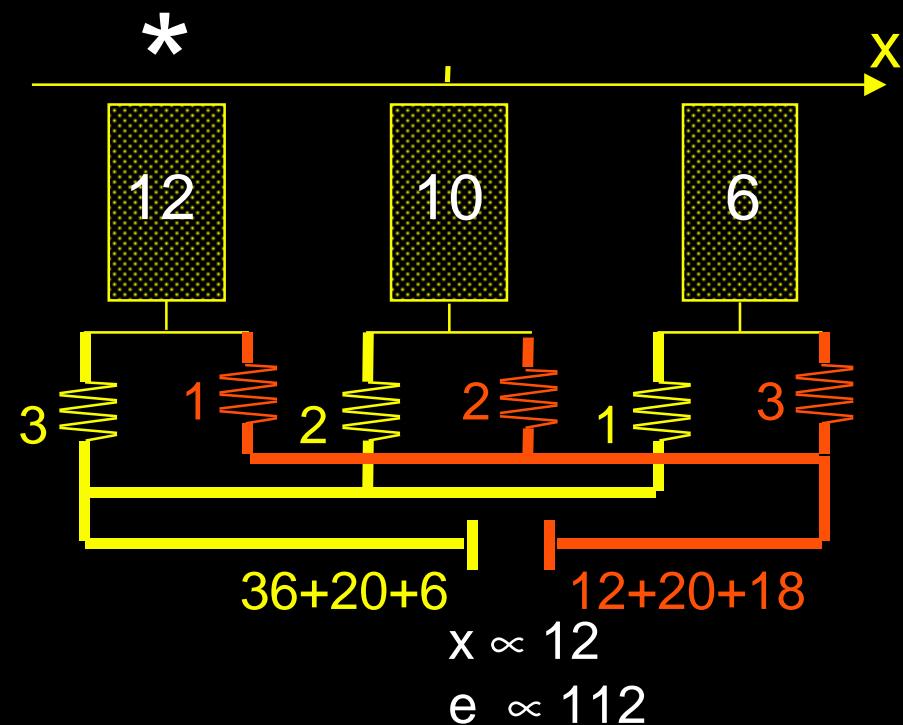
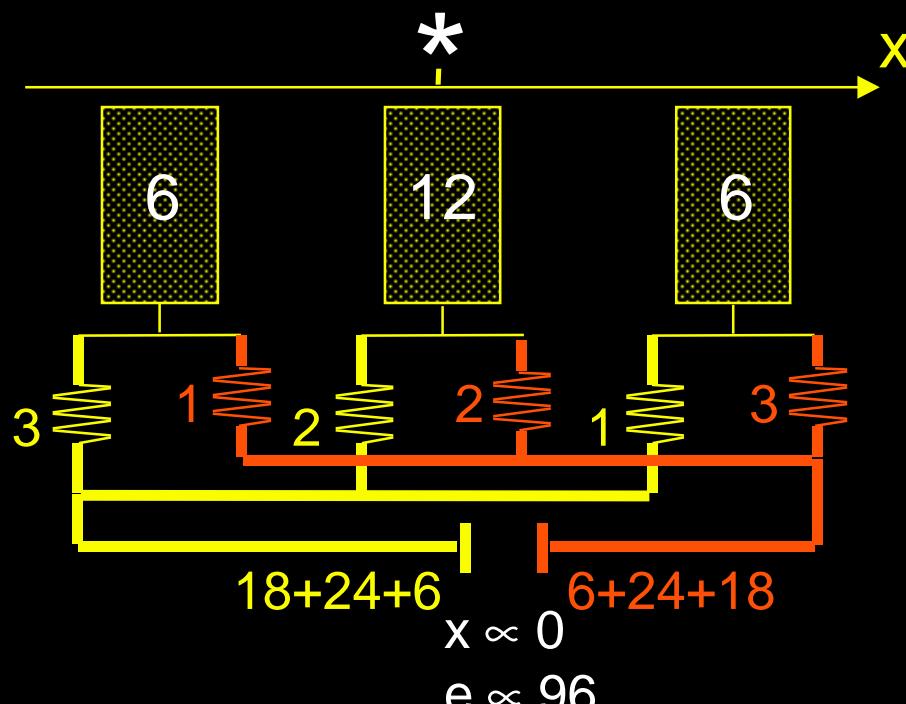
Circuit de positionnement



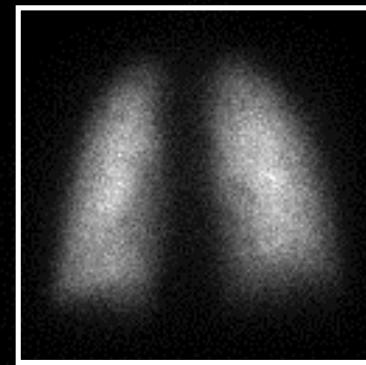
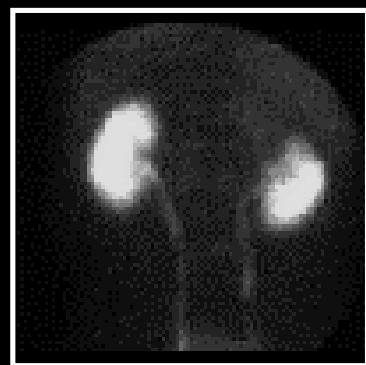
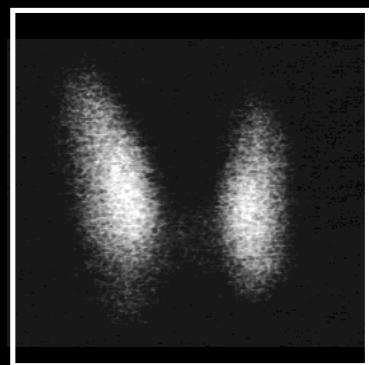
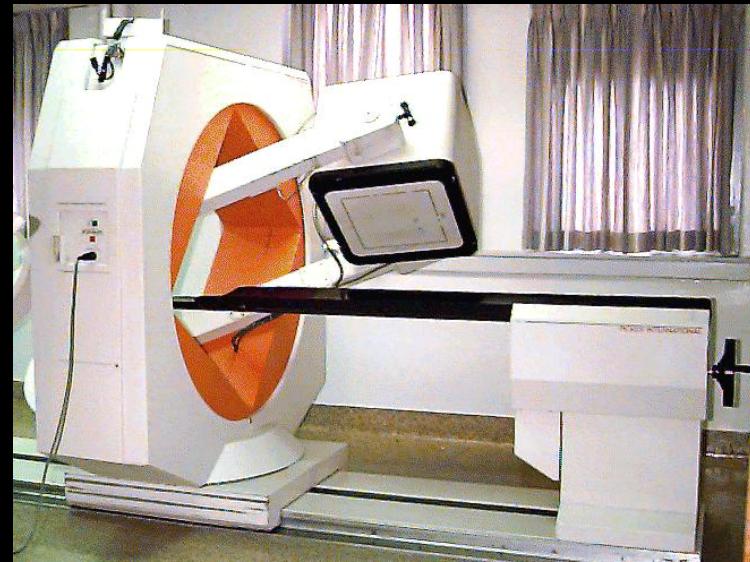
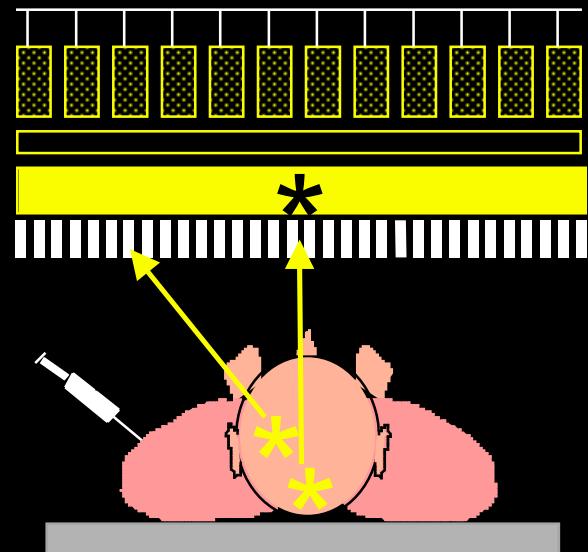
tubes PM

guide de lumière

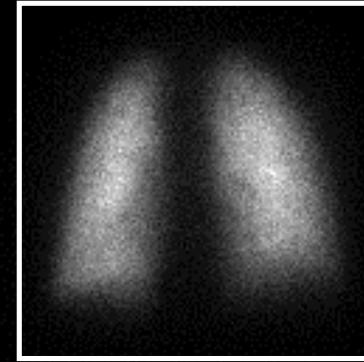
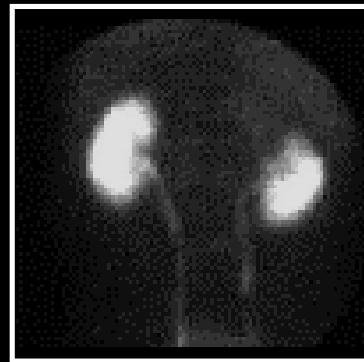
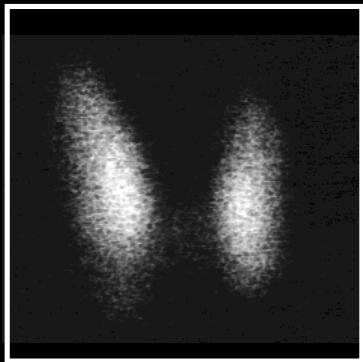
cristal



Images planaires

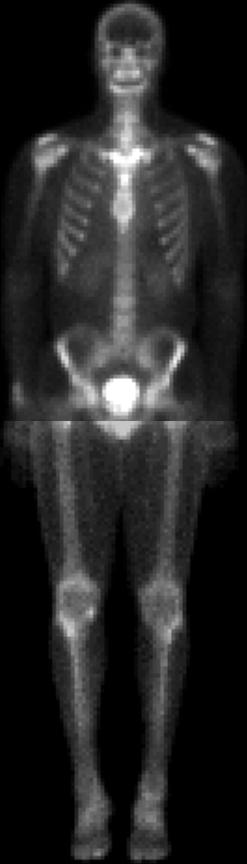


Caractéristiques de l'imagerie monophotonique

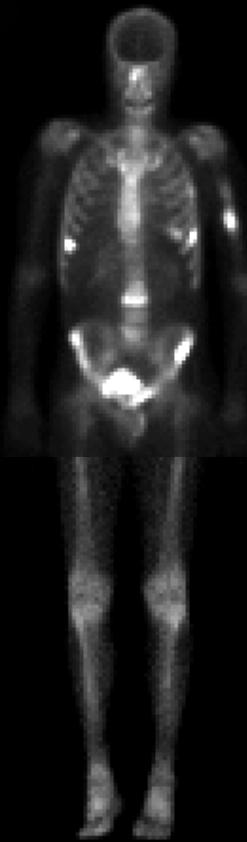
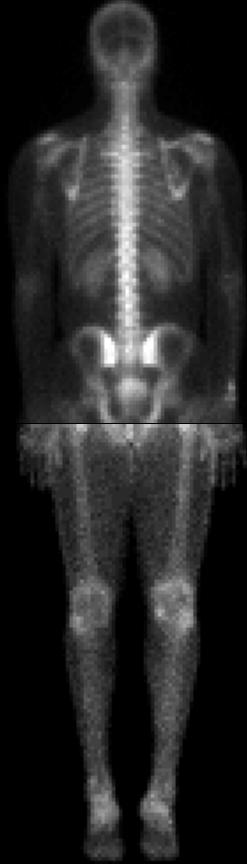


- résolution spatiale entre 8 et 12 mm
- durée d'un examen : entre 10 min et > 1 h

Scintigraphie osseuse au Tc-99m



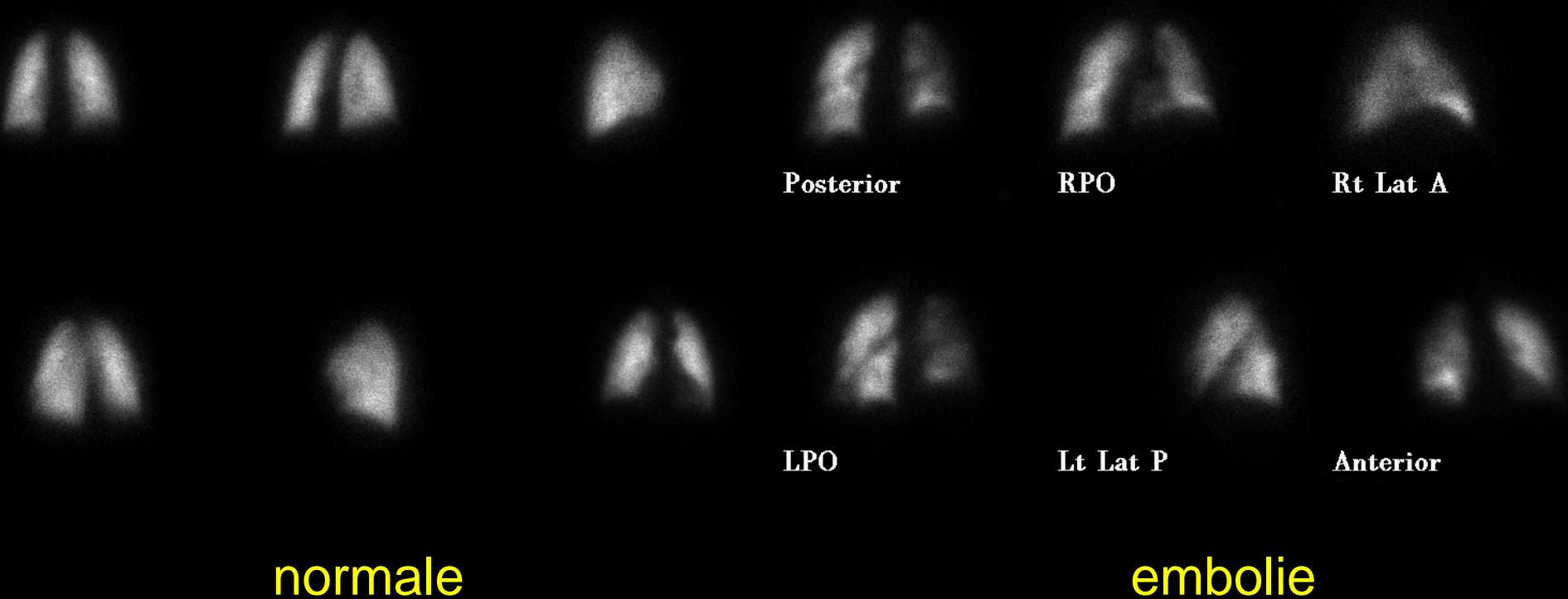
normale



métastases osseuses

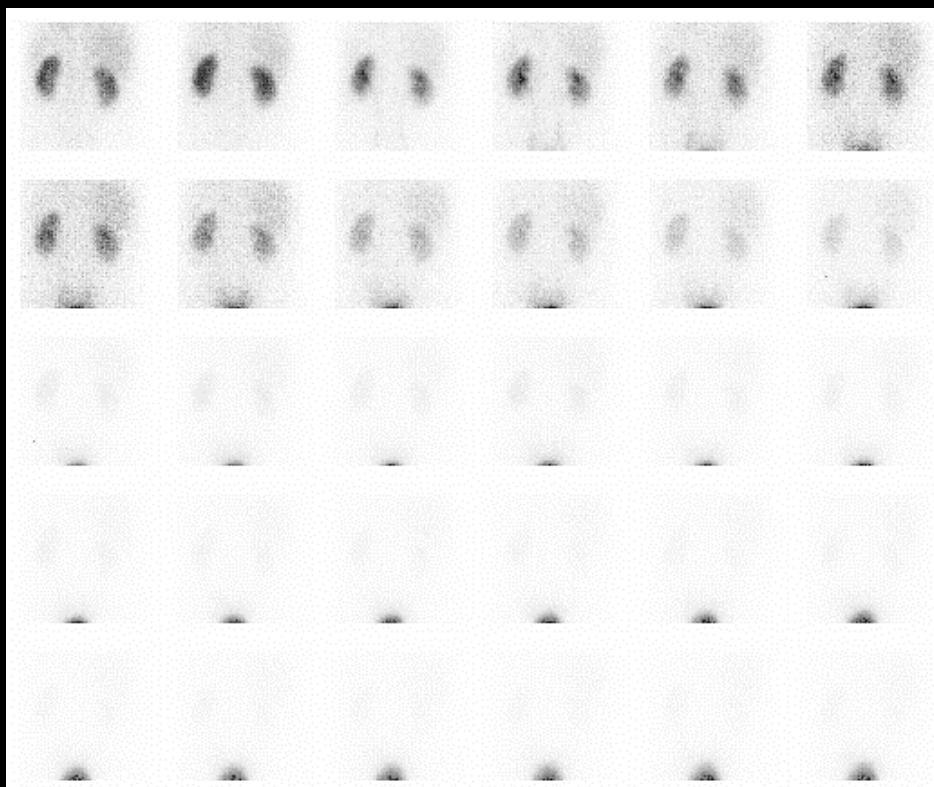


Perfusion pulmonaire : Tc-99m MAA

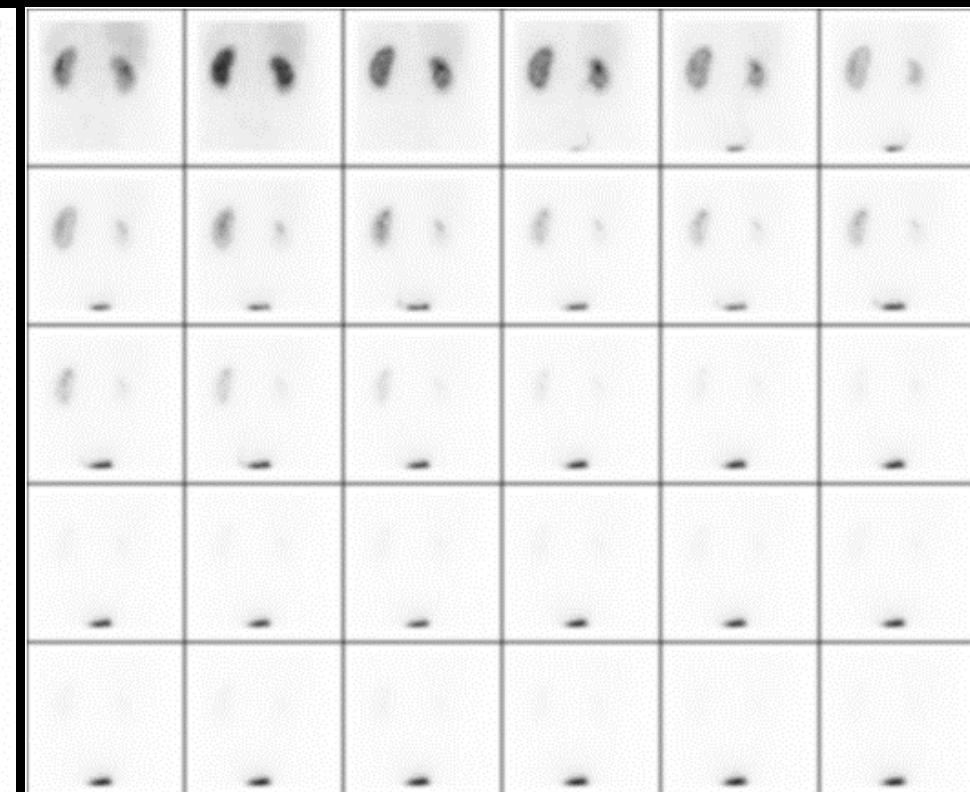


Imagerie dynamique

Tc-99m



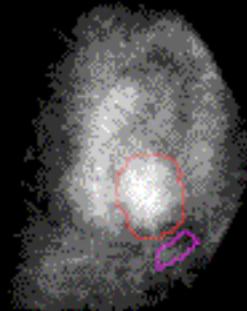
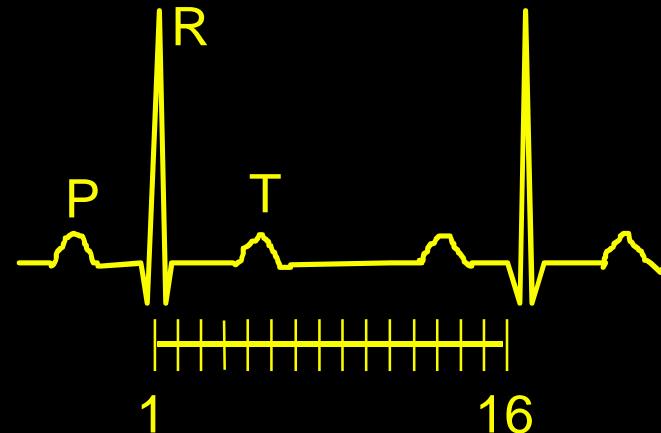
avant captopril



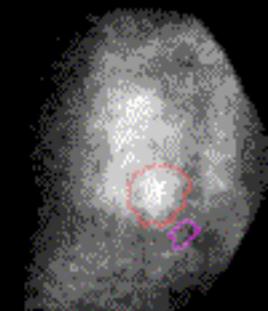
après captopril

Durée totale de l'étude : ~ 3 h

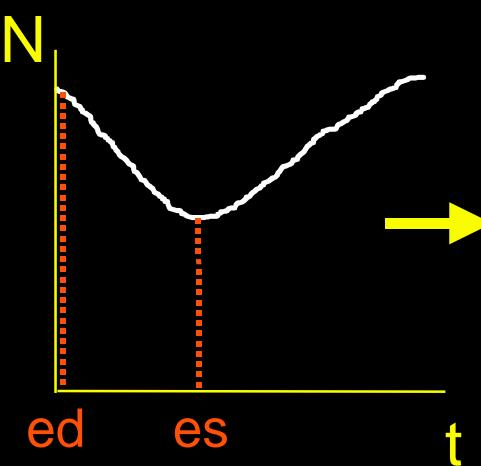
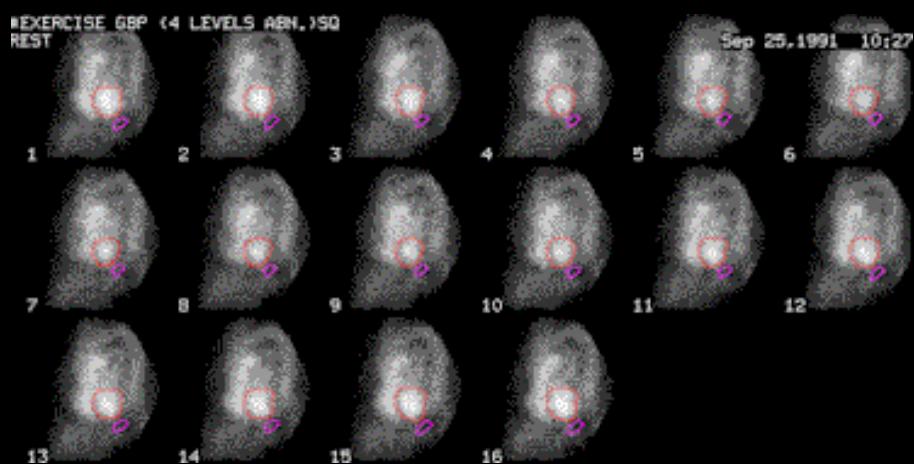
Imagerie cardiaque dynamique : scintigraphie cavitaire Tc-99m



fin de diastole



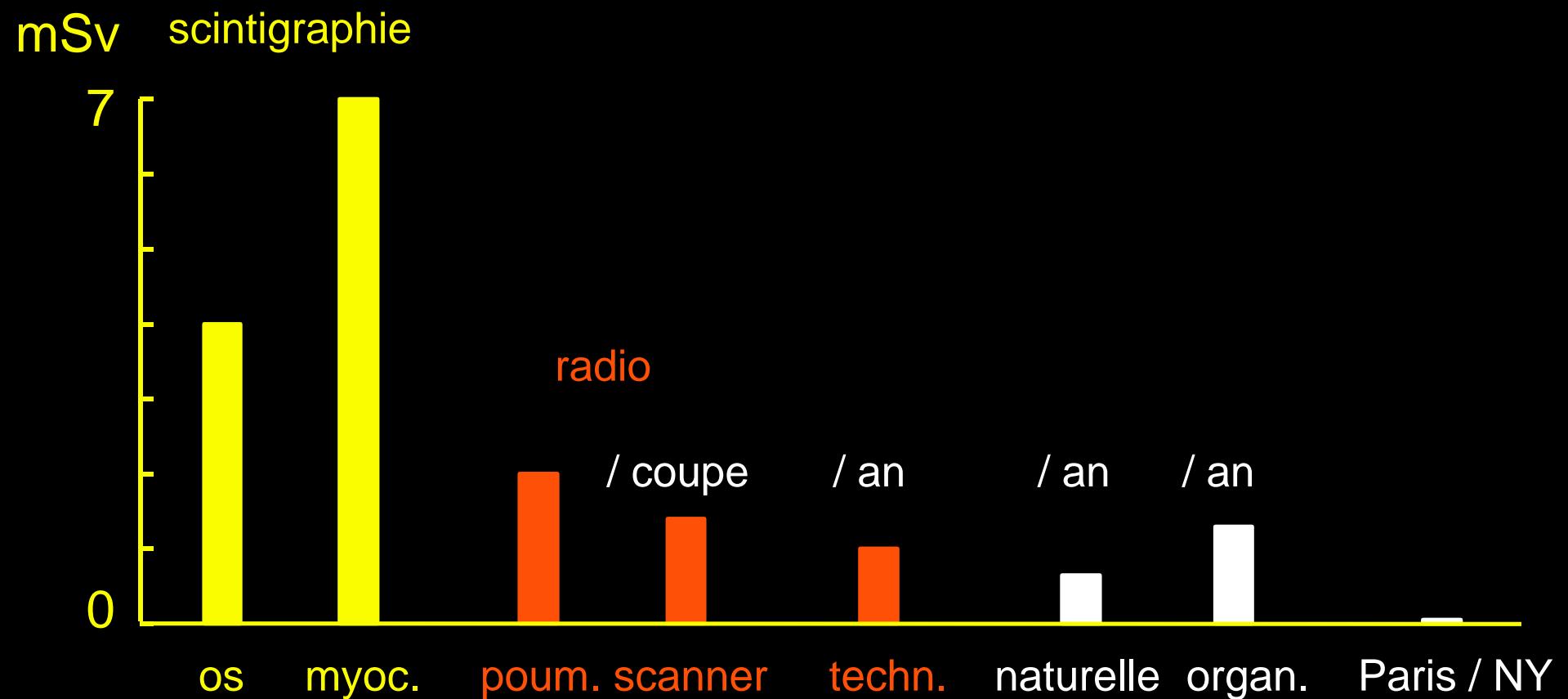
fin de systole



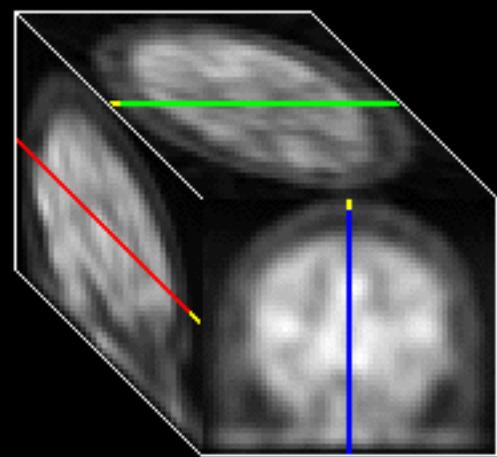
fraction d'éjection

$$100 \times \frac{N_{ed} - N_{es}}{N_{es}}$$

Irradiations liées aux examens



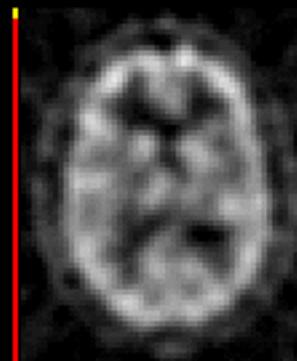
Imagerie planaire (2D) → Imagerie tomographique (3D)



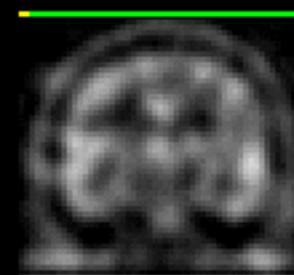
projections 2D



Sagittal



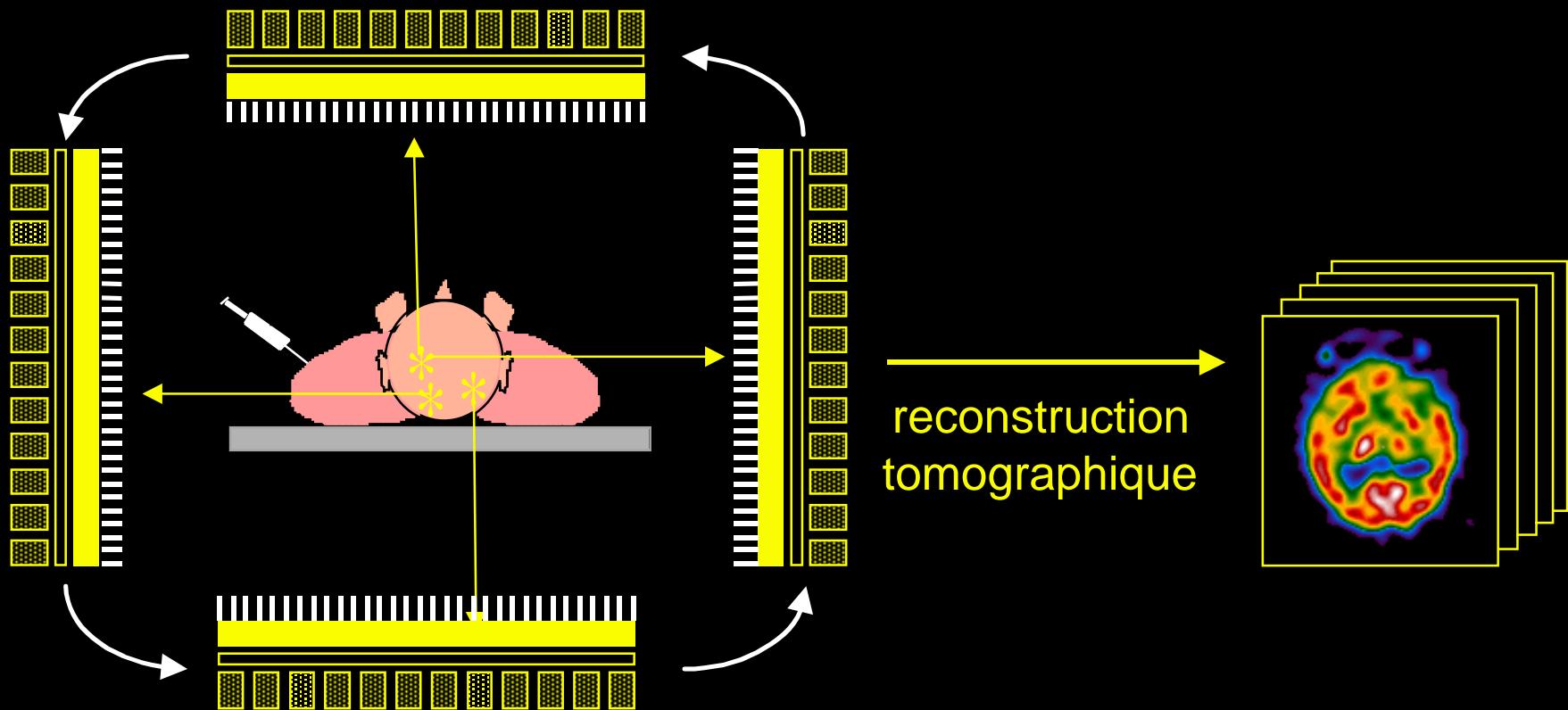
Transverse



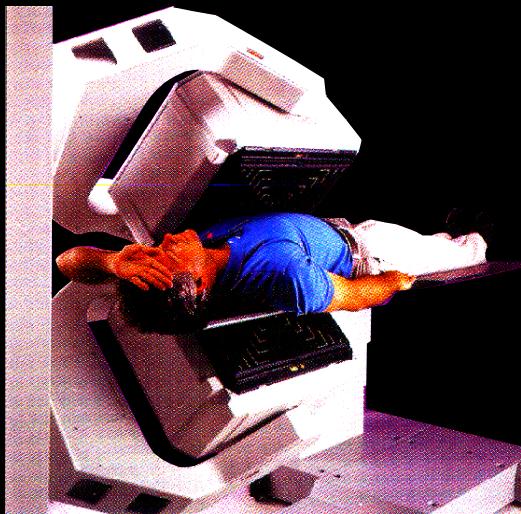
Coronal

3D

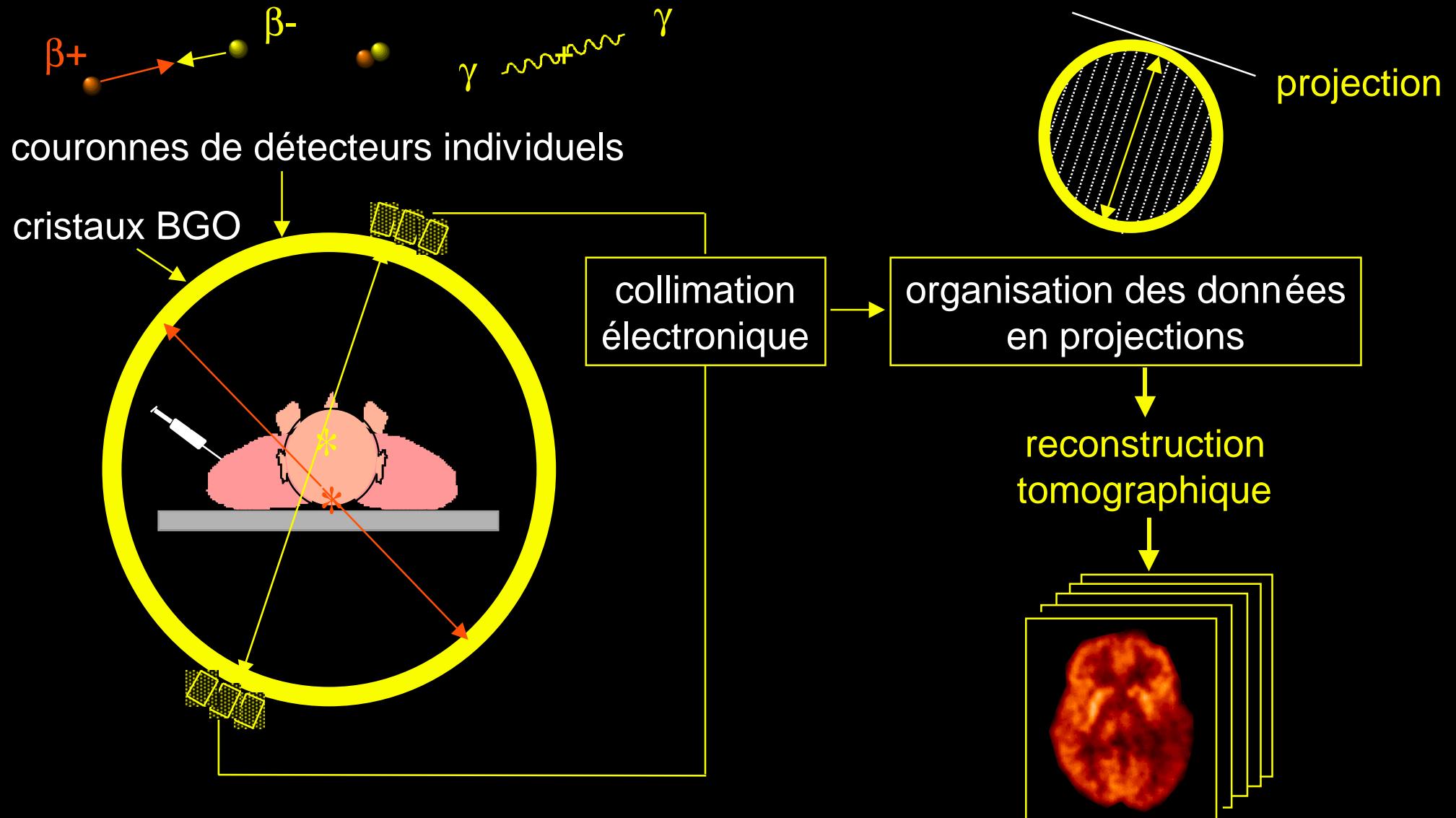
Principe de la tomographie d'émission monophotonique (SPECT)



Différents types de détecteurs en SPECT

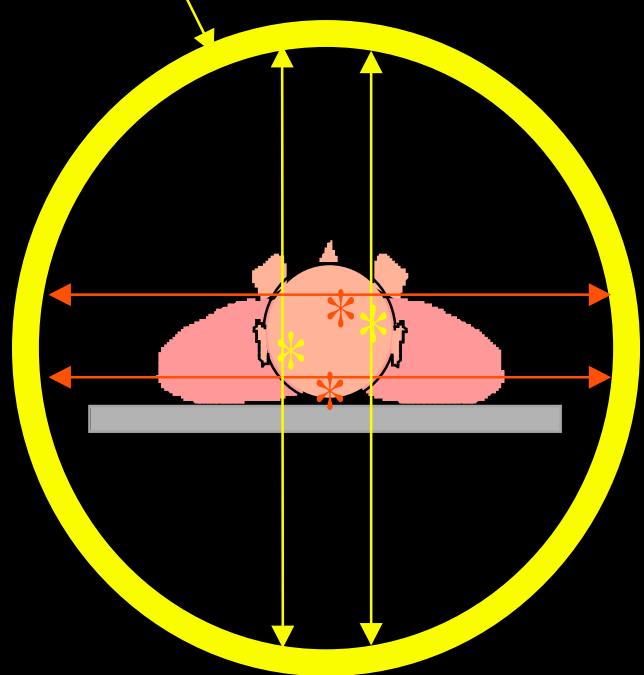


Détecteur pour la tomographie d'émission de positons (PET)



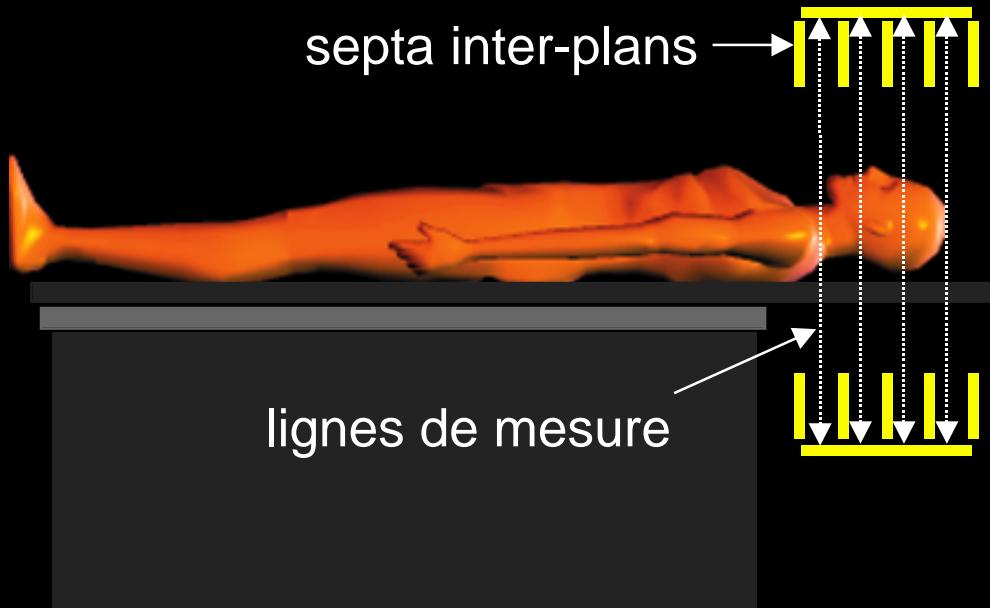
PET bidimensionnel (2D)

couronnes de détecteurs



Vue transaxiale

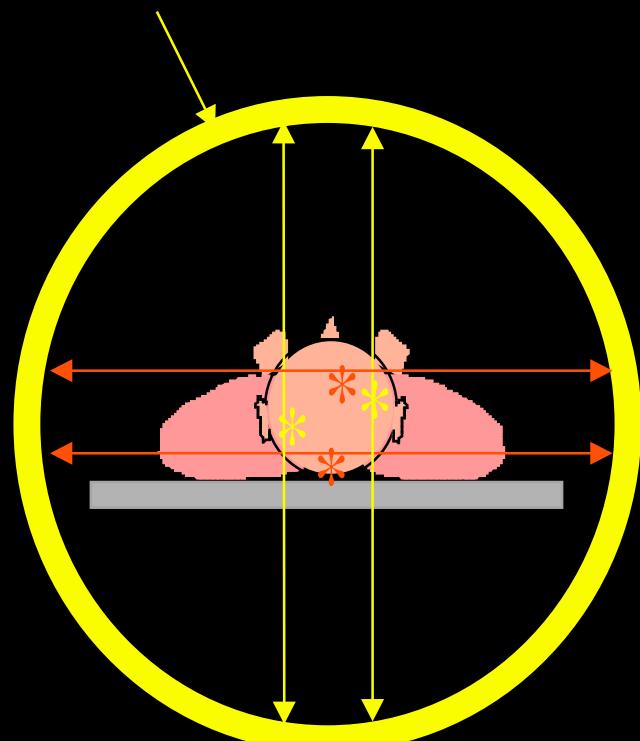
septa inter-plans →



Vue axiale

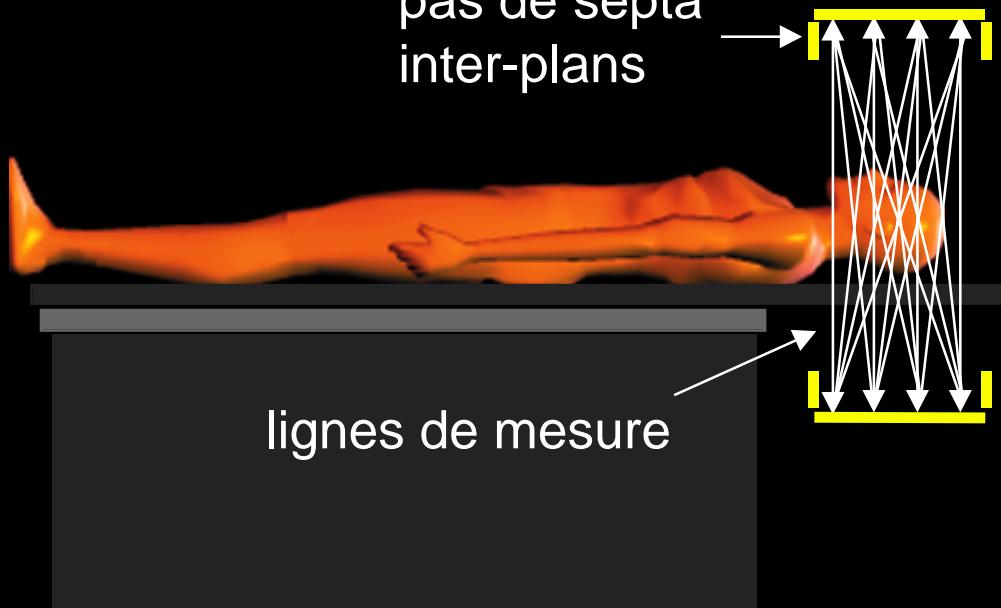
PET tridimensionnel (3D)

couronnes de détecteurs



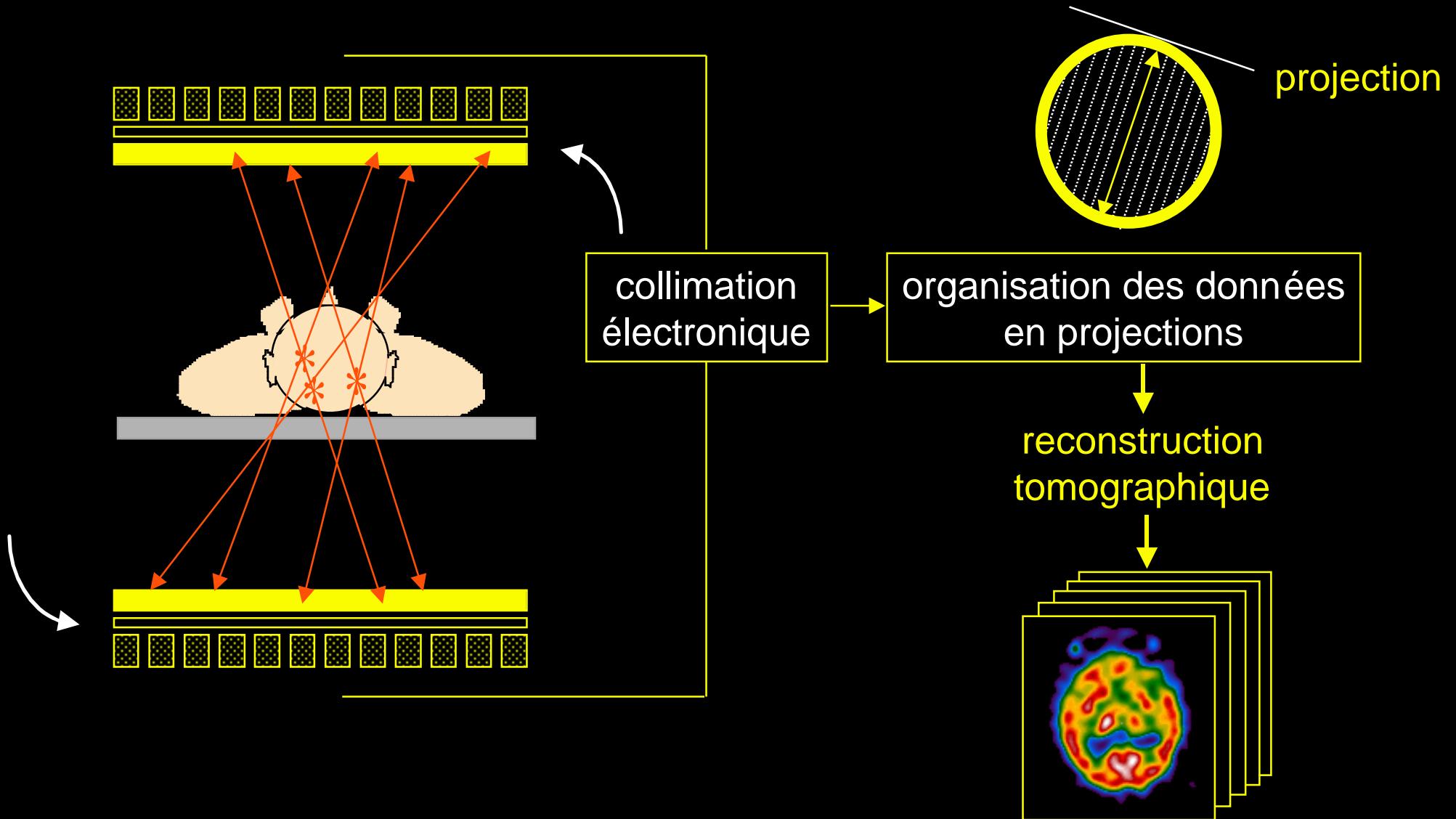
Vue transaxiale

pas de septa
inter-plans



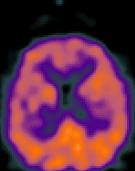
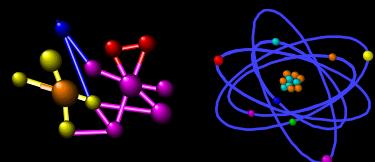
Vue axiale

Détection des positons avec des gamma-caméras



Le traitement de l'information

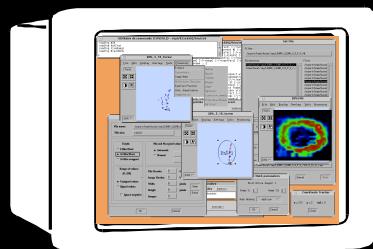
Radiotraceur



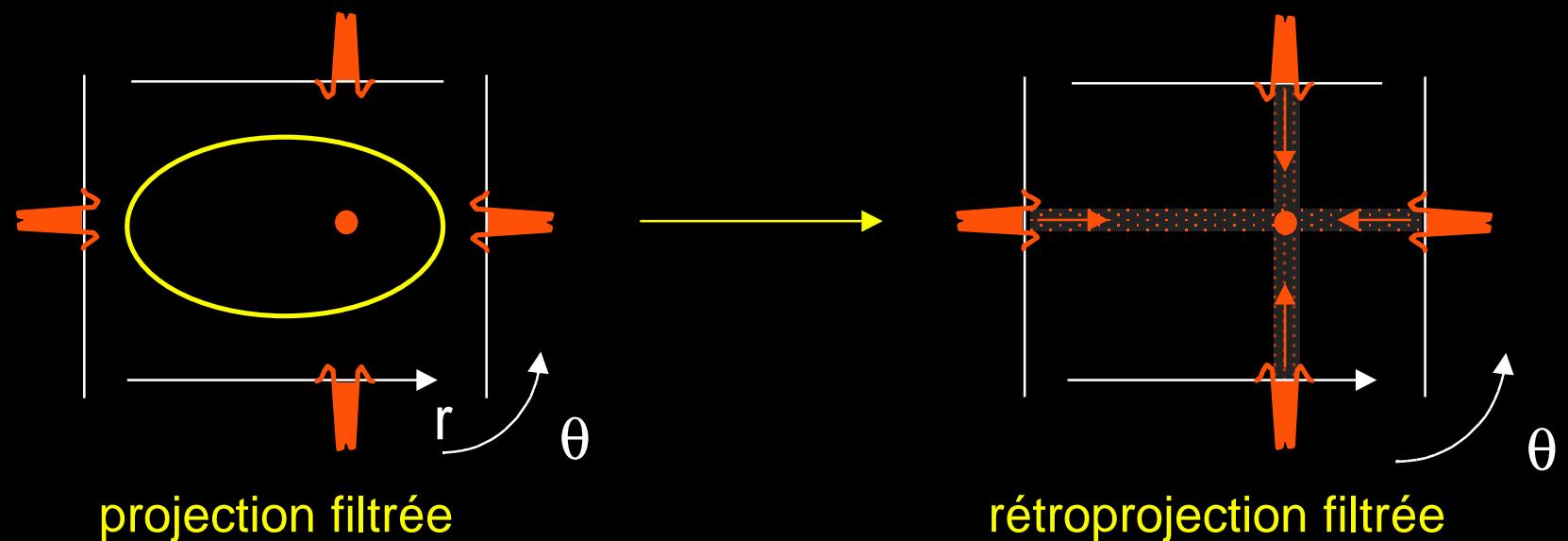
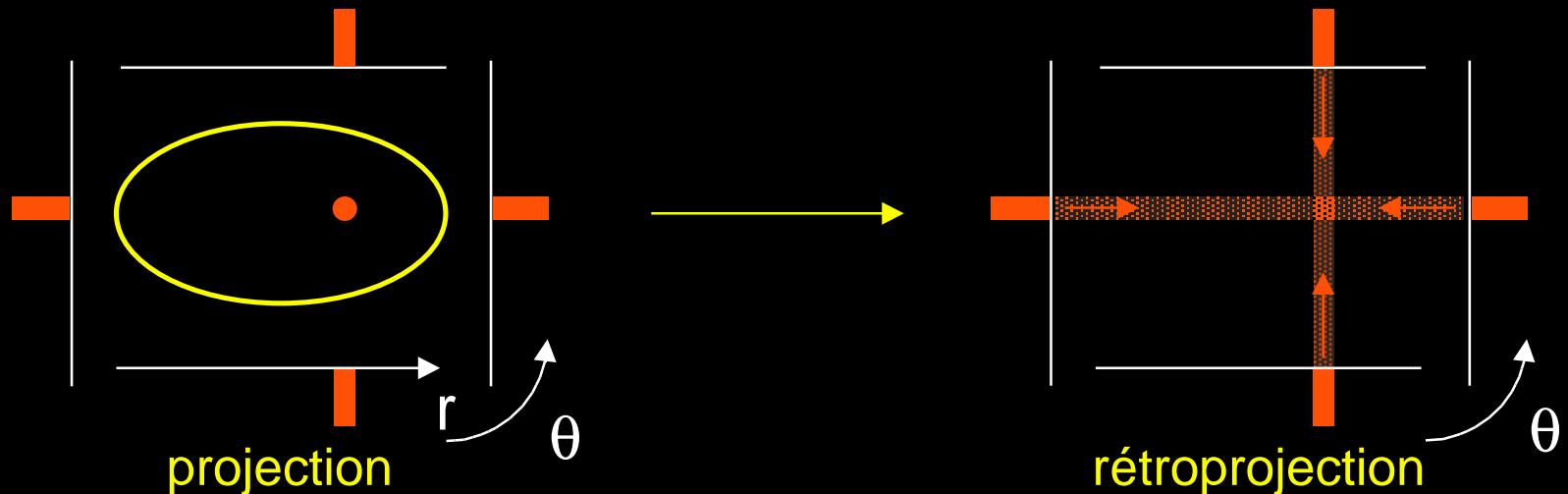
Détecteur



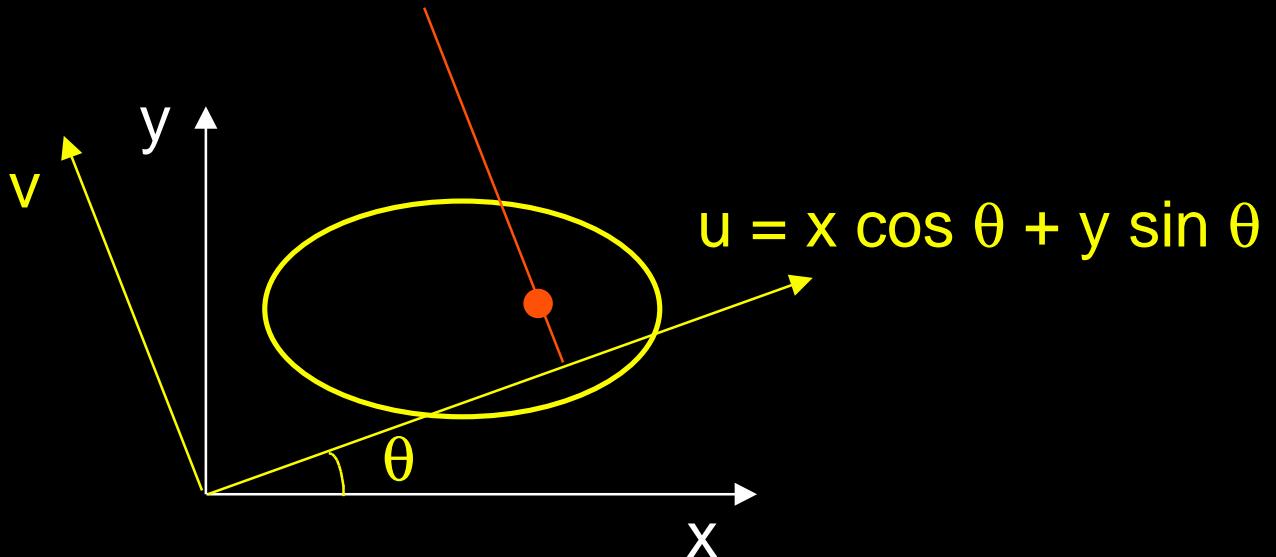
Traitement de l'information



Principe de la reconstruction tomographique



Opérateurs impliqués en reconstruction tomographique



projection

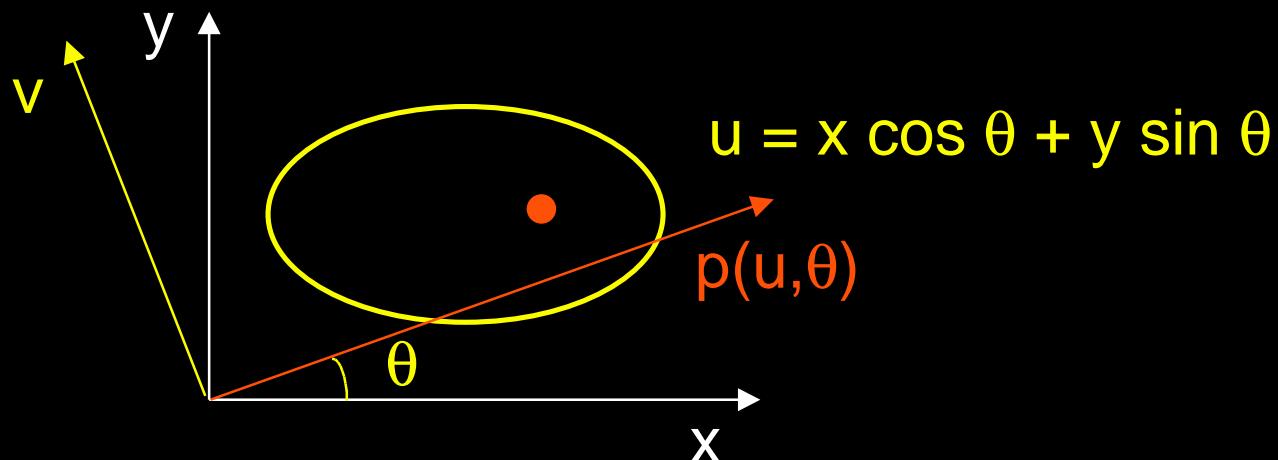
$$p(u, \theta) = \int_{-\infty}^{+\infty} f(x, y) \, dv$$

rétroprojection

$$f^*(x, y) = \int_0^\pi p(u, \theta) \, d\theta$$

Théorème de la tranche centrale

$$p(u,\theta) = \int_{-\infty}^{+\infty} f(x,y) dv \xrightarrow{\text{transformée de Fourier}} P(\rho,\theta) = \int_{-\infty}^{+\infty} p(u,\theta) e^{-i2\pi\rho u} du$$



$$\begin{aligned}\rho_x &= \rho \cos \theta \\ \rho_y &= \rho \sin \theta \\ du.dv &= dx.dy\end{aligned}$$

$$P(\rho,\theta) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} f(x,y) e^{-i2\pi\rho u} du.dv = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} f(x,y) e^{-i2\pi(x\rho_x + y\rho_y)} dx dy$$

Rétroprojection filtrée

$$P(\rho, \theta) = F(\rho_x, \rho_y)$$

$$f(x, y) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} F(\rho_x, \rho_y) e^{i2\pi(x\rho_x + y\rho_y)} d\rho_x d\rho_y$$

$$= \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} P(\rho, \theta) e^{i2\pi(x\rho_x + y\rho_y)} d\rho_x d\rho_y$$

$$= \int_0^{\pi} \int_{-\infty}^{+\infty} P(\rho, \theta) |\rho| e^{i2\pi\rho u} d\rho d\theta$$

$$= \int_0^{\pi} p'(u, \theta) d\theta \quad \text{avec} \quad p'(u, \theta) = \int_{-\infty}^{+\infty} P(\rho, \theta) |\rho| e^{i2\pi\rho u} d\rho$$

$$\rho_x = \rho \cos \theta \\ \rho_y = \rho \sin \theta$$

$$\rho = \sqrt{\rho_x^2 + \rho_y^2} \\ d\rho_x d\rho_y = \rho d\rho d\theta \\ u = x \cos \theta + y \sin \theta$$

Algorithme de rétroprojection filtrée

$$f(x,y) = \int_0^{\pi} p'(u, \theta) d\theta \quad \text{avec} \quad p' (u, \theta) = \int_{-\infty}^{+\infty} P(\rho, \theta) |\rho| e^{i2\pi\rho u} d\rho$$

projections

$$p(u, \theta)$$

TF

images reconstruites

$$f(x,y)$$

rétroprojection

$$P(\rho, \theta)$$

filtrage

$$|\rho| P(\rho, \theta)$$

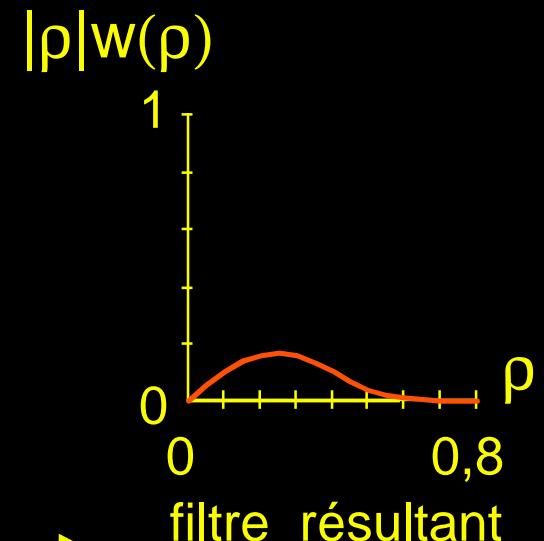
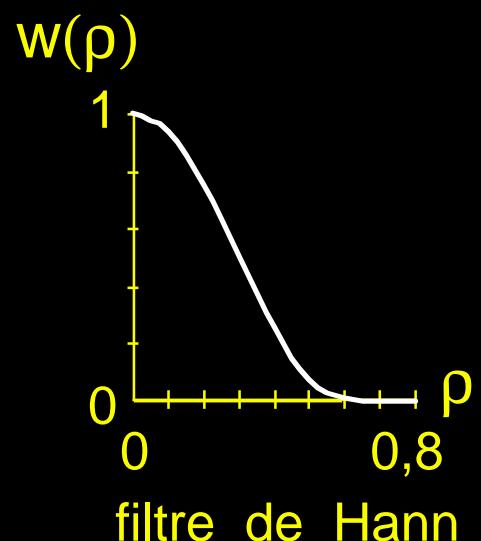
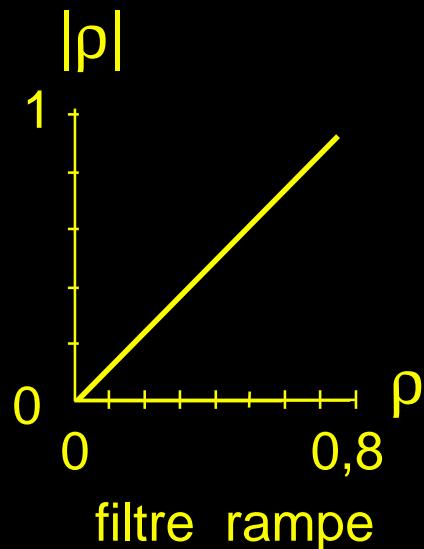
TF⁻¹

$$p'(u, \theta)$$

Insuffisance du filtre rampe

$$f(x,y) = \int_0^{\pi} p'(u, \theta) d\theta \quad \text{avec} \quad p' (u, \theta) = \int_{-\infty}^{+\infty} P(\rho, \theta) |\rho| e^{i 2\pi \rho u} d\rho$$

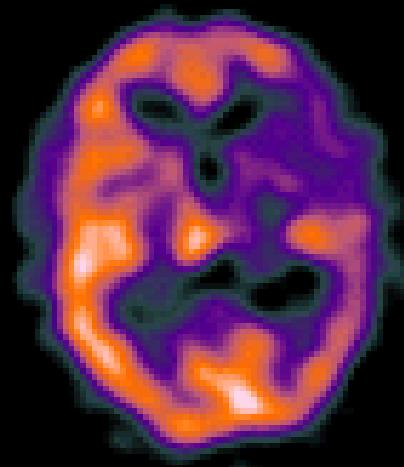
$|\rho| \longrightarrow |\rho|w(\rho)$



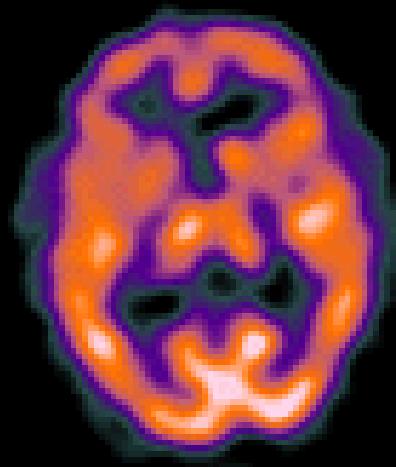
$$w(\rho) = 0,5 \times (1 + \cos \pi \rho / \rho_c) \quad \begin{matrix} \text{si } \rho < \rho_c \\ = 0 \end{matrix} \quad \longleftrightarrow \quad 1/16 \begin{pmatrix} 1 & 2 & 1 \\ 2 & 4 & 2 \\ 1 & 2 & 1 \end{pmatrix} \quad \begin{matrix} \text{si } \rho_c = \rho_N \\ \text{si } \rho > \rho_c \end{matrix}$$

SPECT cérébral HMPAO Tc-99m

Syndrome de fatigue chronique



Avant traitement

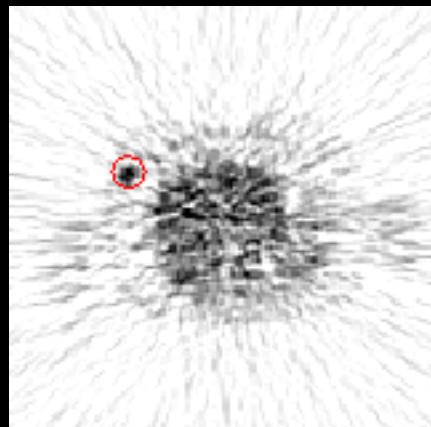


Après traitement

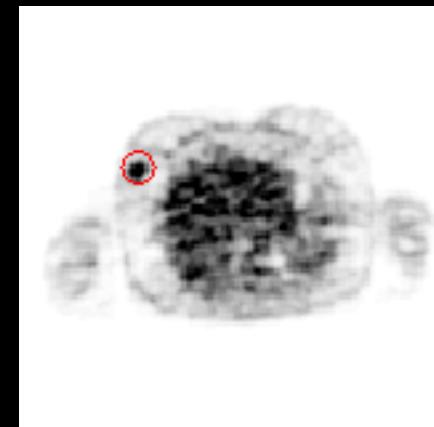


IRM anatomique

Reconstruction analytique vs reconstruction algébrique

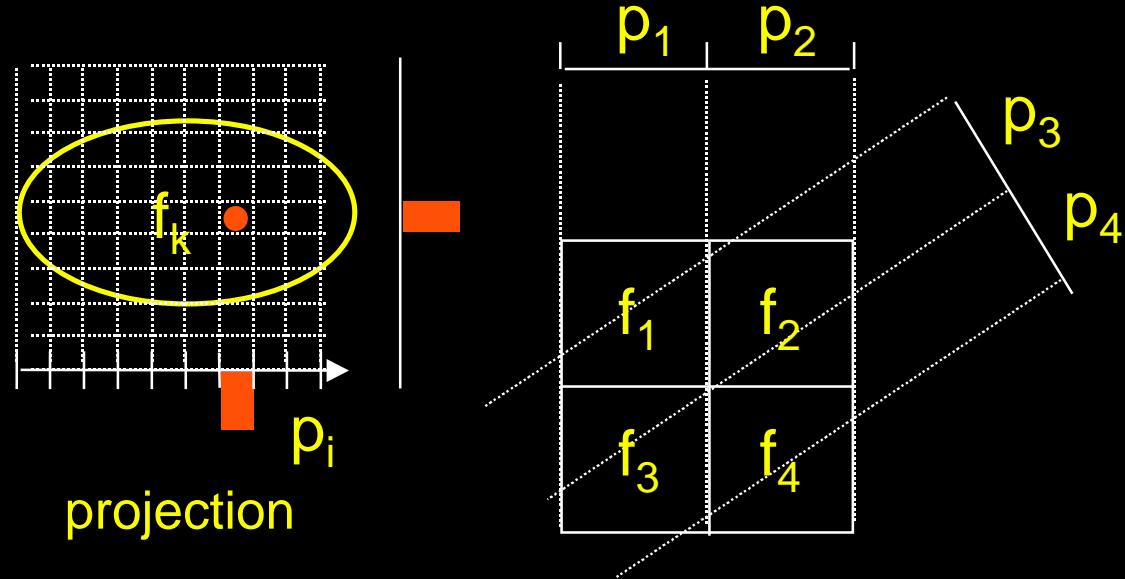


rétroprojection filtrée



reconstruction algébrique

Principe des méthodes de reconstruction algébrique



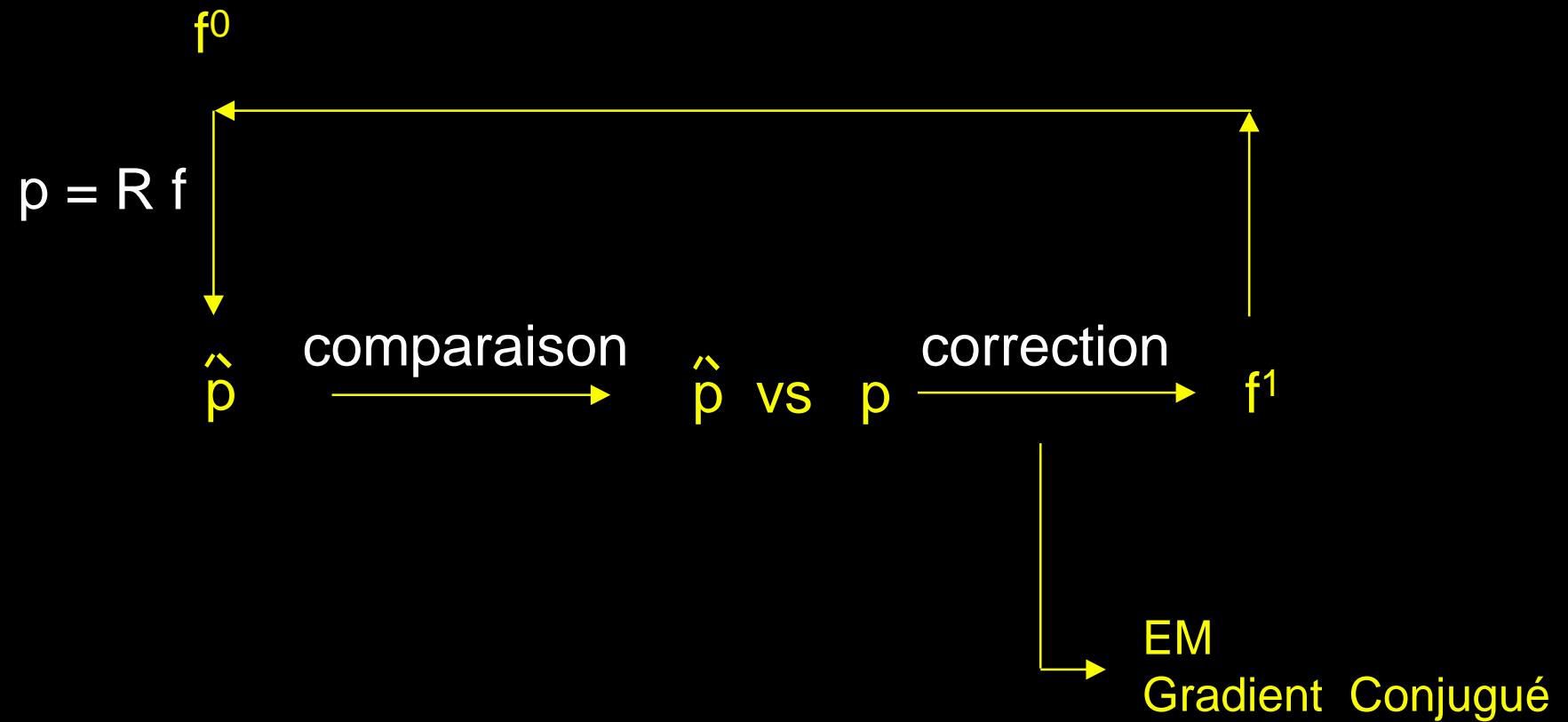
$$\begin{aligned} p_1 &= r_{11} f_1 + r_{12} f_2 + r_{13} f_3 + r_{14} f_4 \\ p_2 &= r_{21} f_1 + r_{22} f_2 + r_{23} f_3 + r_{24} f_4 \\ p_3 &= r_{31} f_1 + r_{32} f_2 + r_{33} f_3 + r_{34} f_4 \\ p_4 &= r_{41} f_1 + r_{42} f_2 + r_{43} f_3 + r_{44} f_4 \end{aligned}$$

$$\begin{bmatrix} p_1 \\ p_2 \\ p_3 \\ p_4 \end{bmatrix} = \begin{bmatrix} r_{11} & \dots & r_{14} \\ \vdots & \ddots & \vdots \\ r_{41} & \dots & r_{44} \end{bmatrix} \begin{bmatrix} f_1 \\ f_2 \\ f_3 \\ f_4 \end{bmatrix}$$

$$p = R f$$

Résolution du problème inverse

$$p = R f$$



Algorithmes de reconstruction itérative

EM
(Expectation Maximization)

$$f^{n+1} = f^n R^t \frac{p}{R f^n}$$

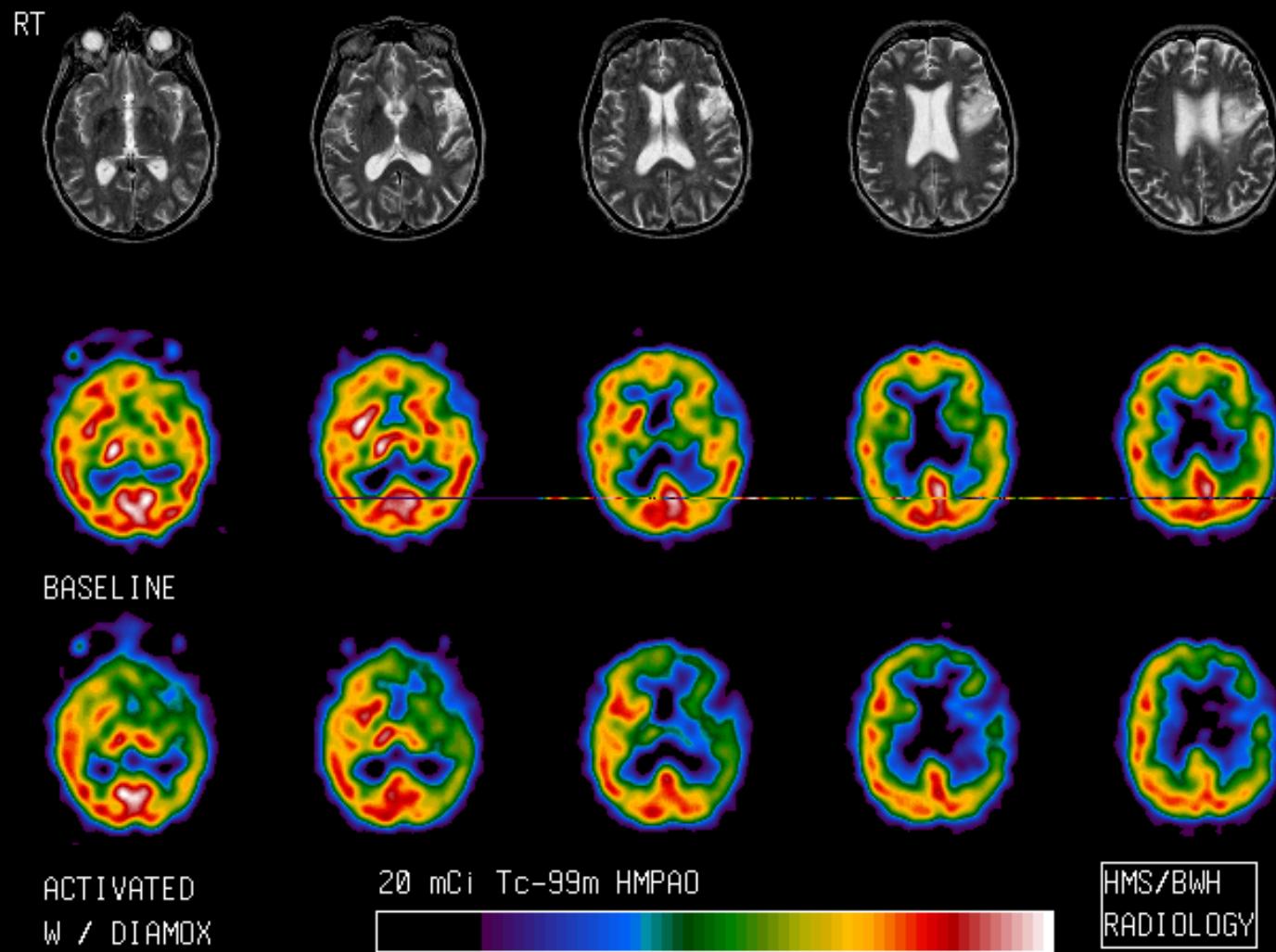
solution positive
convergence lente

Gradient Conjugué

$$f^{n+1} = f^n + \alpha^n d^n$$

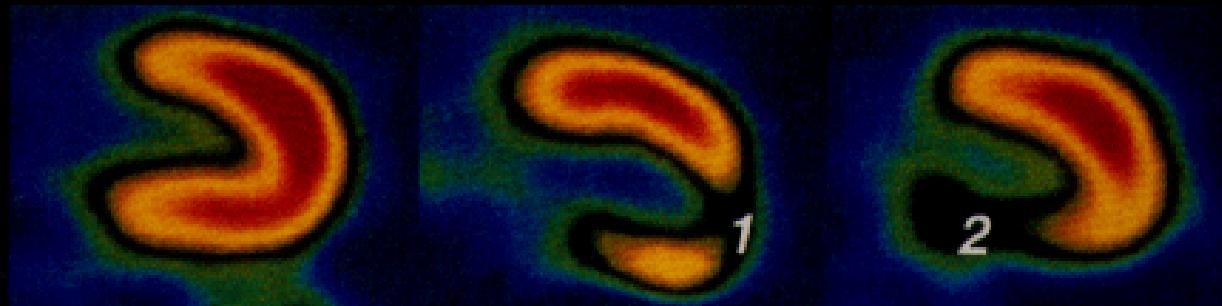
$$\| p - R f \| ^2$$

SPECT cérébral HMPAO Tc-99m

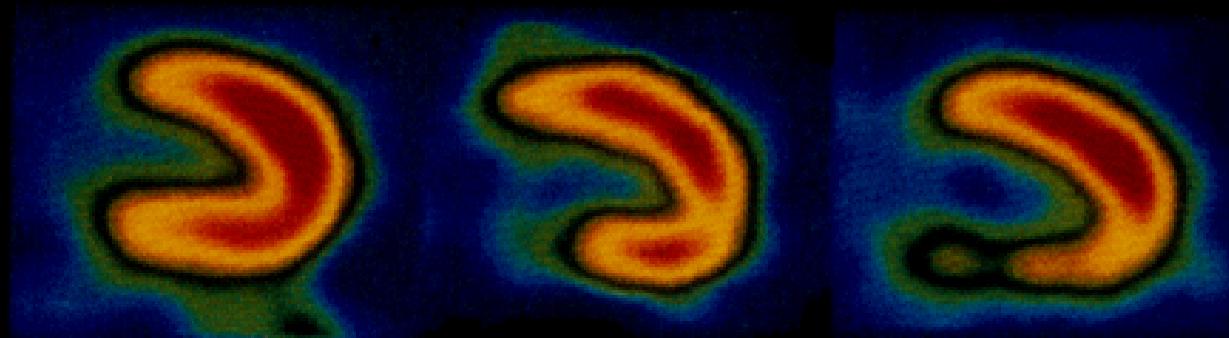


Tomoscintigraphie cardiaque au Tl-201

effort



repos



normal

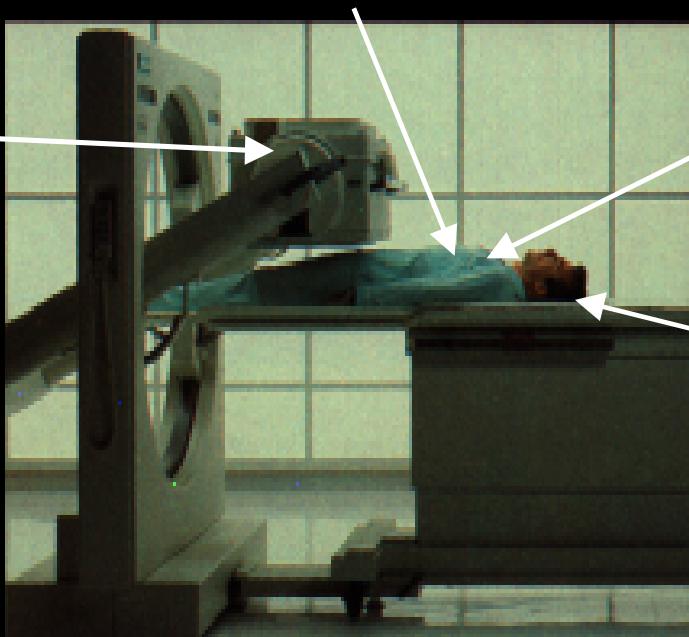
ischémie

infarctus

Vers la quantification ?

Technologiques
uniformité
résolution spatiale variable
résolution spatiale limitée
bruit de mesure
temps mort

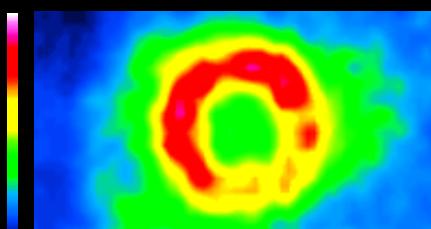
Physiques
atténuation, diffusion, fortuits



Statistiques
émission aléatoire
des photons ou des
positons

Patient
mouvements

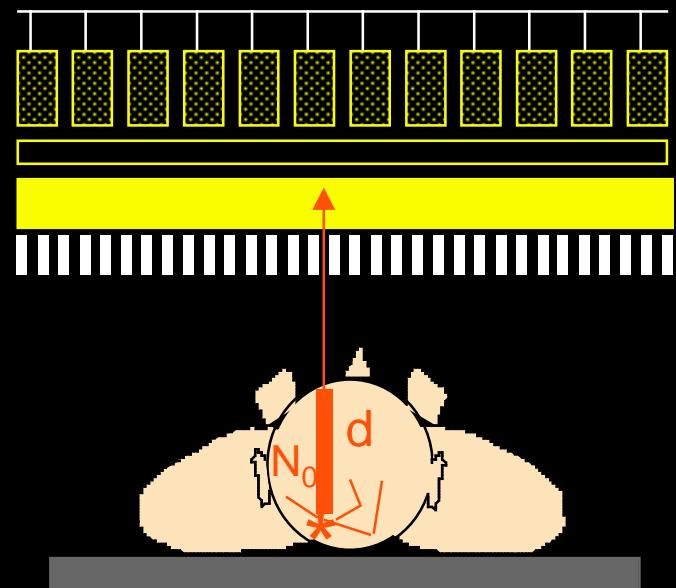
Algorithmiques
reconstruction tomographique



Problème de l'atténuation

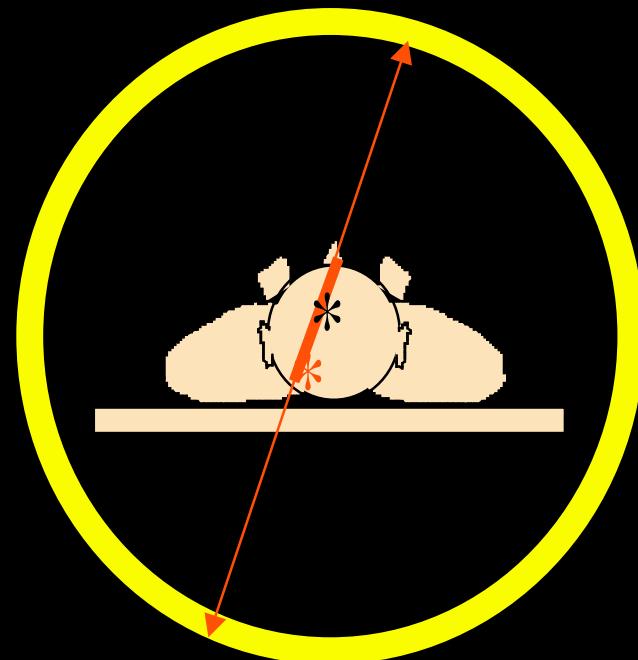
SPECT

$$N = N_0 \exp \int_0^d -\mu(l) dl$$



dépend du lieu d'émission
sur la ligne de projection

PET



ne dépend pas du lieu d'émission
sur la ligne de projection

Importance de l'atténuation

$$N = N_0 \exp \int_0^d -\mu(l) dl$$

0 épaisseur d'atténuation moitié 18 cm



poumons	140 keV
tissus mous	$\mu = 0,04 \text{ cm}^{-1}$
os cortical	$\mu = 0,15 \text{ cm}^{-1}$



0 épaisseur d'atténuation moitié

8 cm

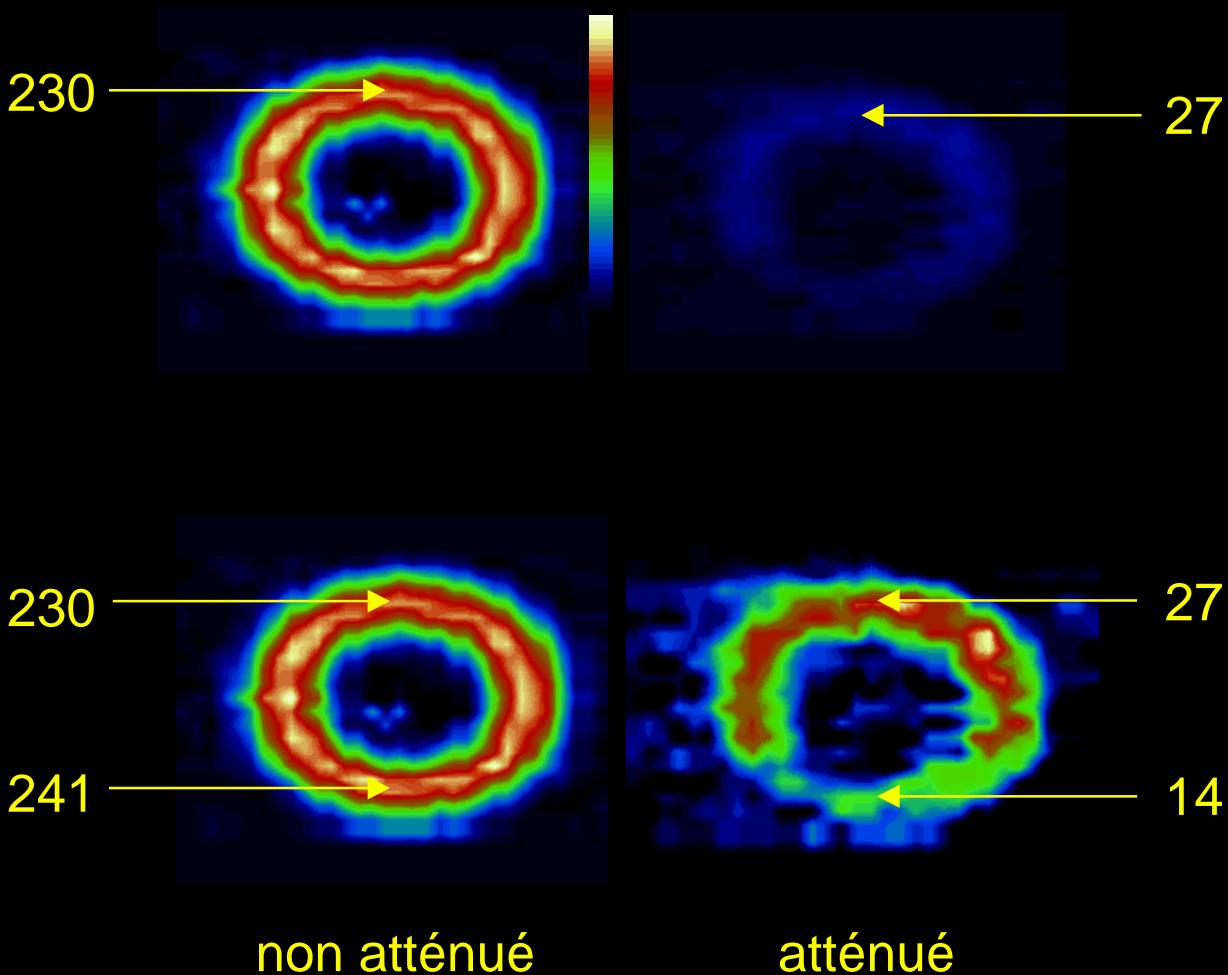


511 keV

Tc-99m 140 keV

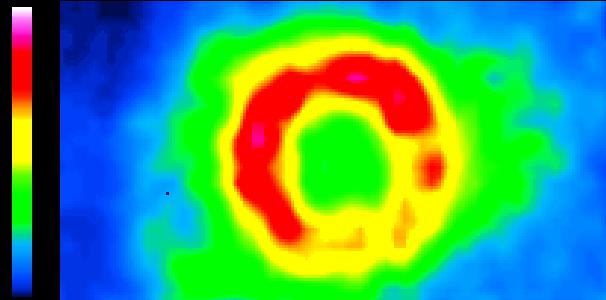
Tl-201 75 keV

Conséquence de l'atténuation en SPECT

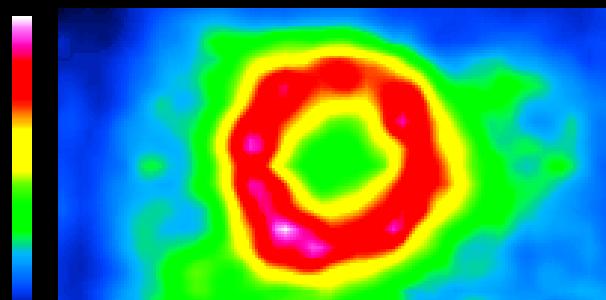


Effet de l'atténuation : exemples cliniques

SPECT TI-201

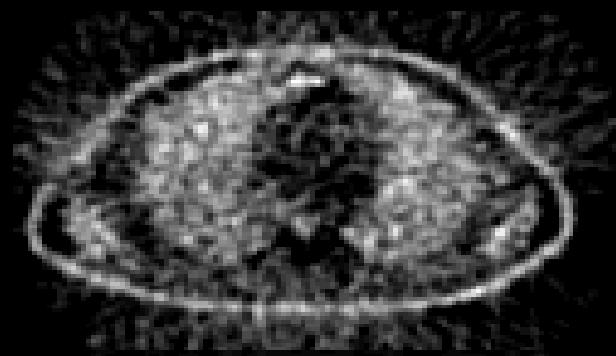


sans correction d'atténuation

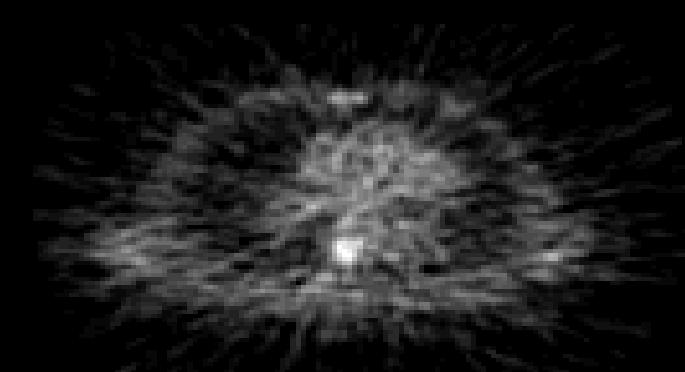


avec correction d'atténuation

PET FDG



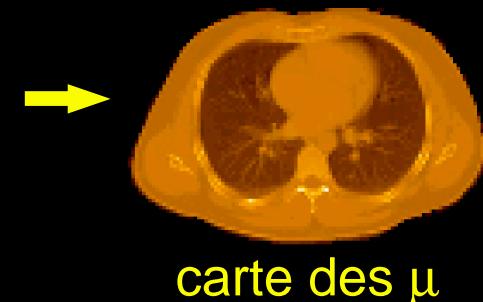
sans correction d'atténuation



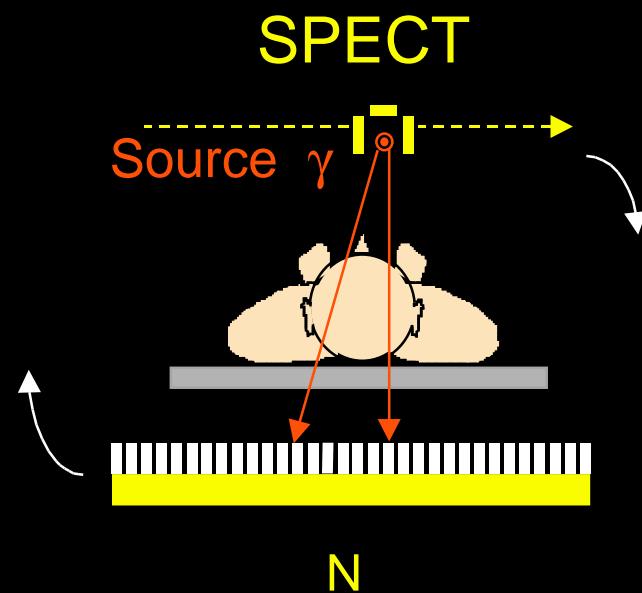
avec correction d'atténuation

Comment compenser de l'atténuation ?

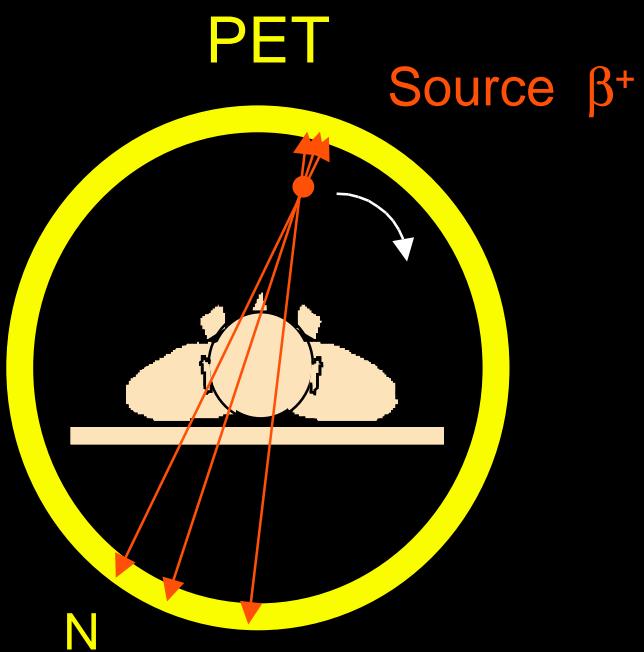
$$N = N_0 \exp \int_0^d -\mu(l) dl \rightarrow \text{mesure de la densité du milieu atténuant}$$



carte des μ

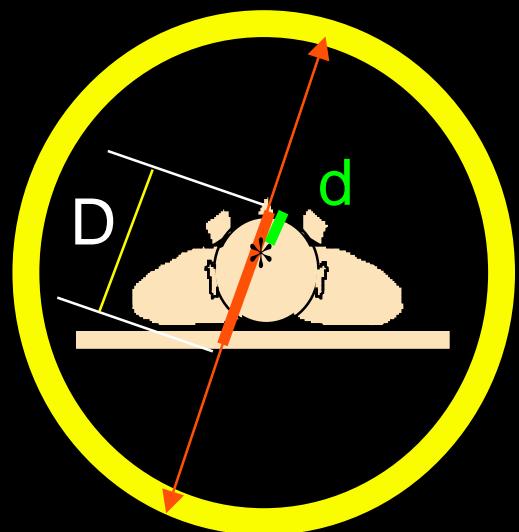


$$\ln \frac{N_0}{N} = \int_0^d \mu(l) dl$$



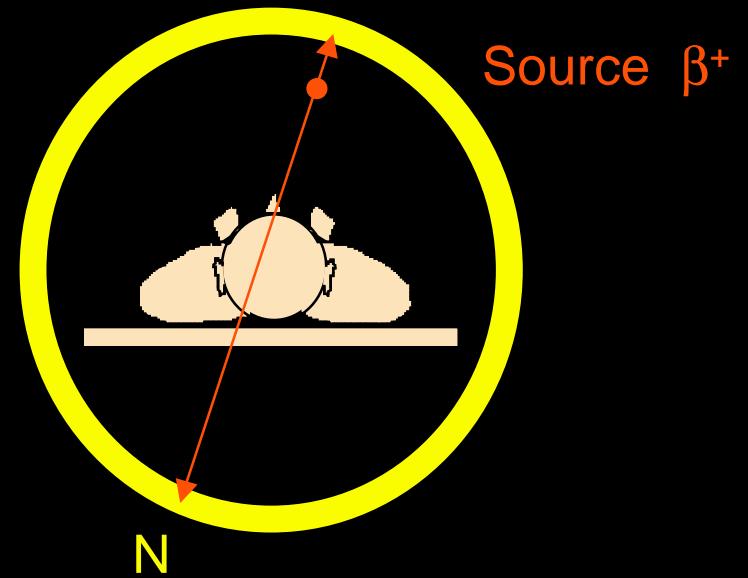
Correction d'atténuation en PET

mesure d'émission



$$N \propto N_0 \exp[-\mu d] \cdot N_0 \exp [-\mu(D-d)] \\ \propto \exp [-\mu D]$$

mesure de transmission

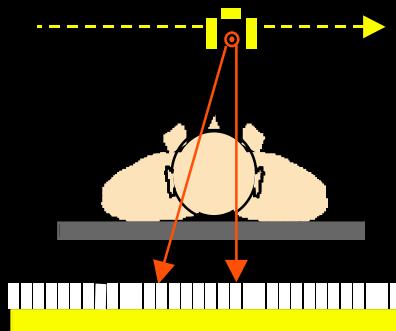


$$\frac{N_0}{N} = \int_0^d \mu(l) dl = \exp [\mu D]$$

nb de coïncidences sans atténuation
= nb de coïncidences détectées $\times N_0 / N$

Correction d'atténuation en SPECT

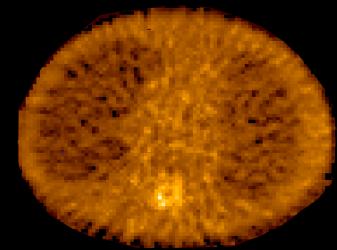
mesure en
transmission



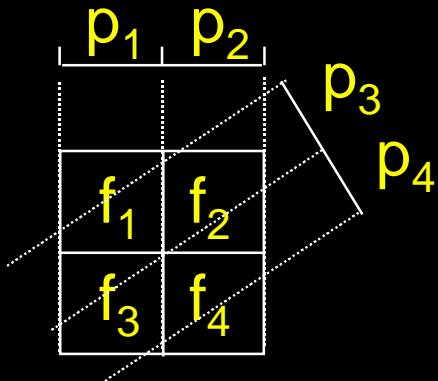
reconstruction
tomographique

$$\ln \frac{N_0}{N} = \int_0^d \mu(l) dl$$

carte des μ

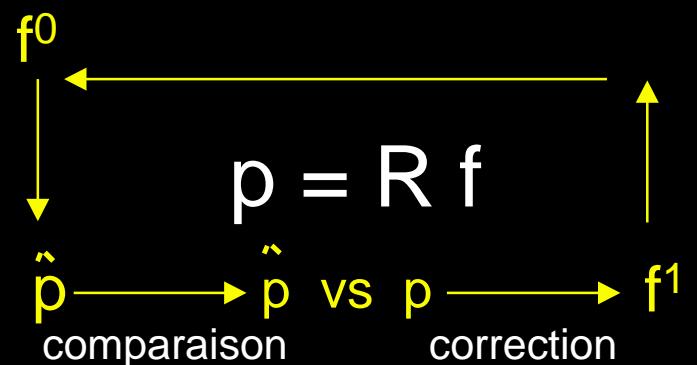


modélisation de l'atténuation dans le projecteur R

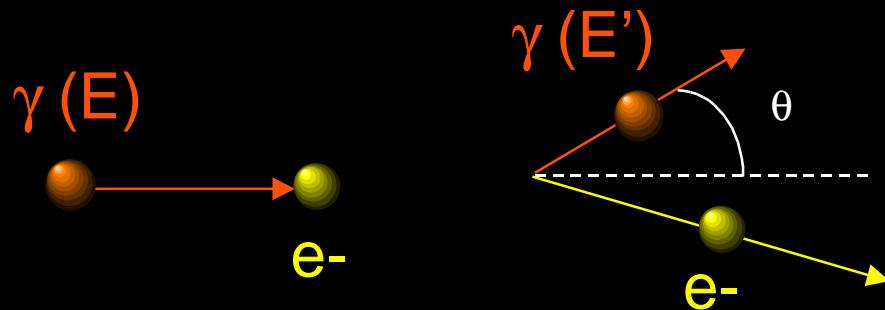


$$p_1 = r_{11}f_1 \exp(-\mu_1 d_1) + r_{12}f_2 \exp(-\mu_2 d_2) + r_{13}f_3 \exp(-\mu_3 d_3) + r_{14}f_4 \exp(-\mu_4 d_4)$$

reconstruction itérative



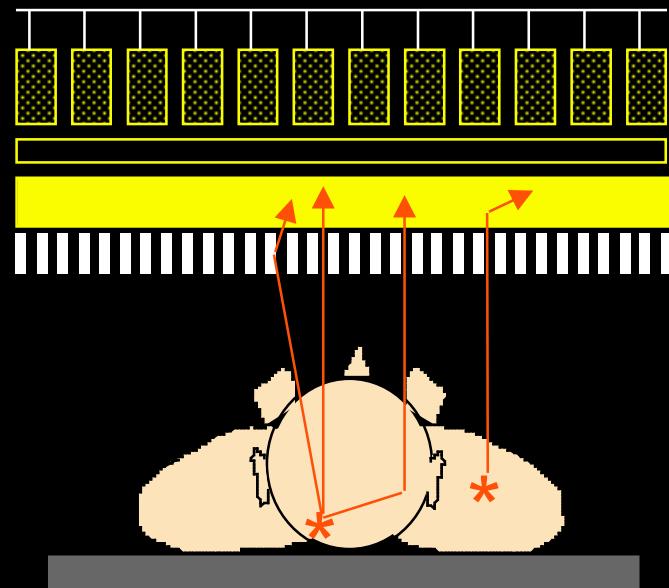
Problème de la diffusion Compton



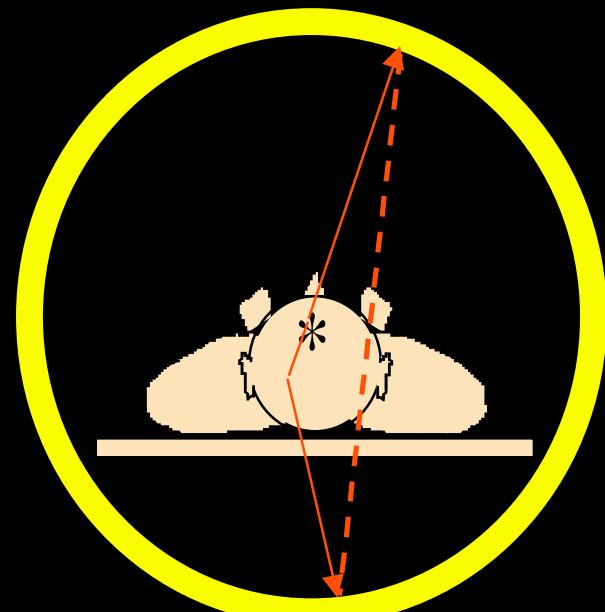
- photons mal localisés
- photons d'énergie diminuée

$$E' = \frac{E}{1 + E(1 - \cos \theta)/m_0 c^2}$$

SPECT

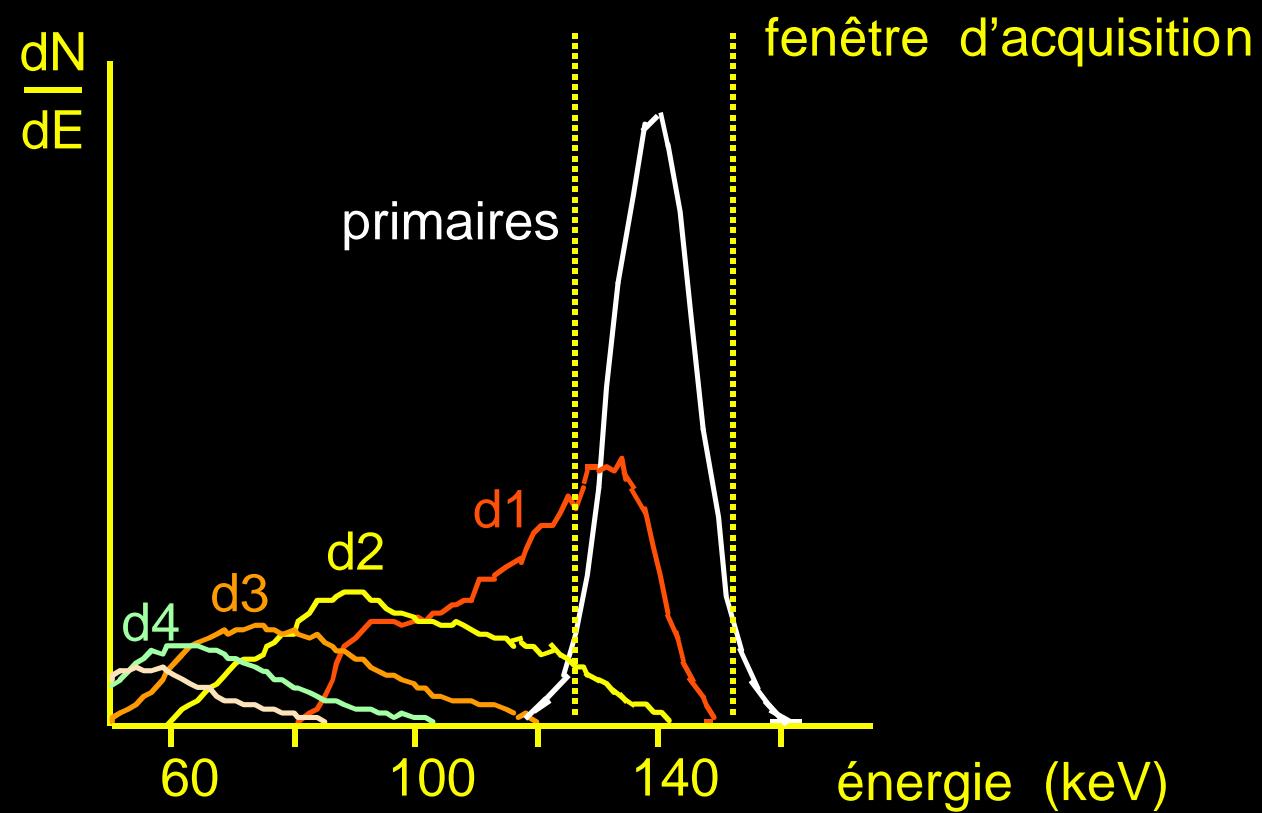


PET



Caractéristiques énergétiques des photons diffusés

Tc-99m (140 keV)



Conséquence de la diffusion en SPECT

Tc-99m

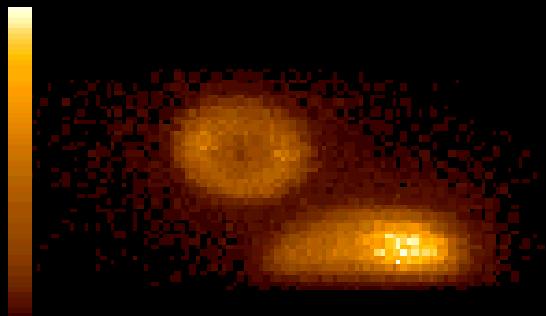
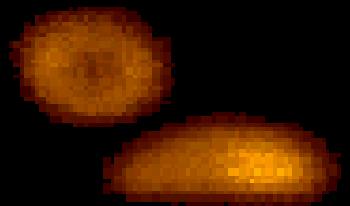
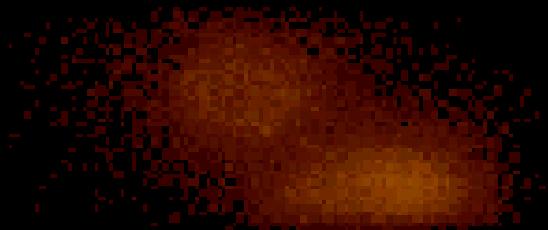


image fenêtre acquisition

=



+

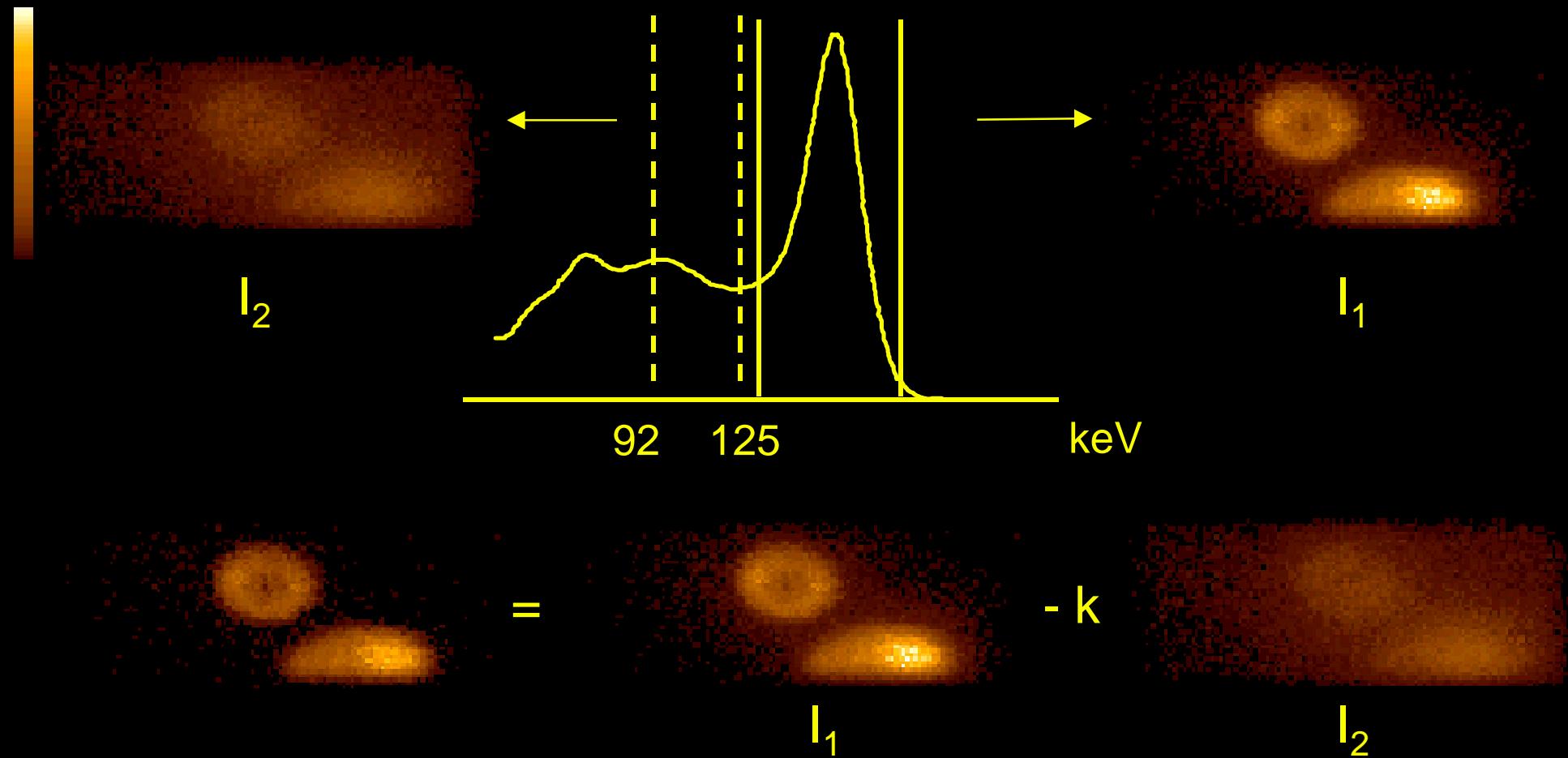


photons primaires

photons diffusés (37%)

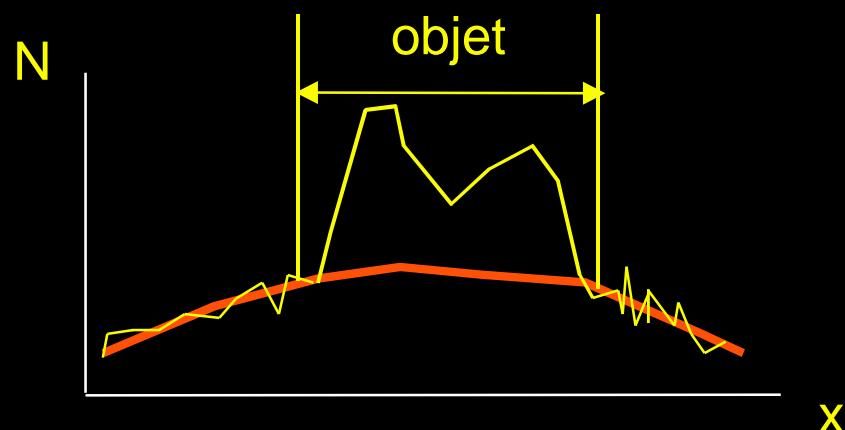
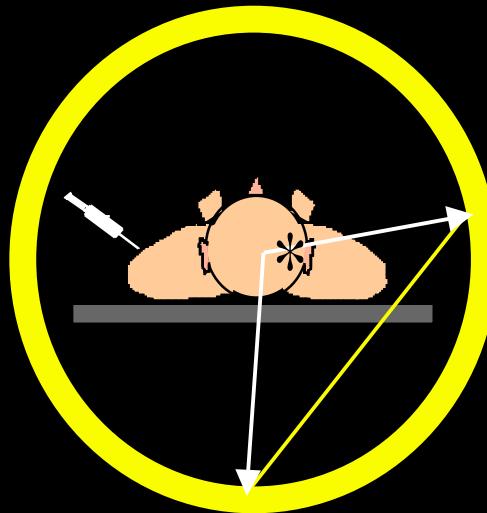
Correction de la diffusion en SPECT ou PET

Tc-99m (140 keV)

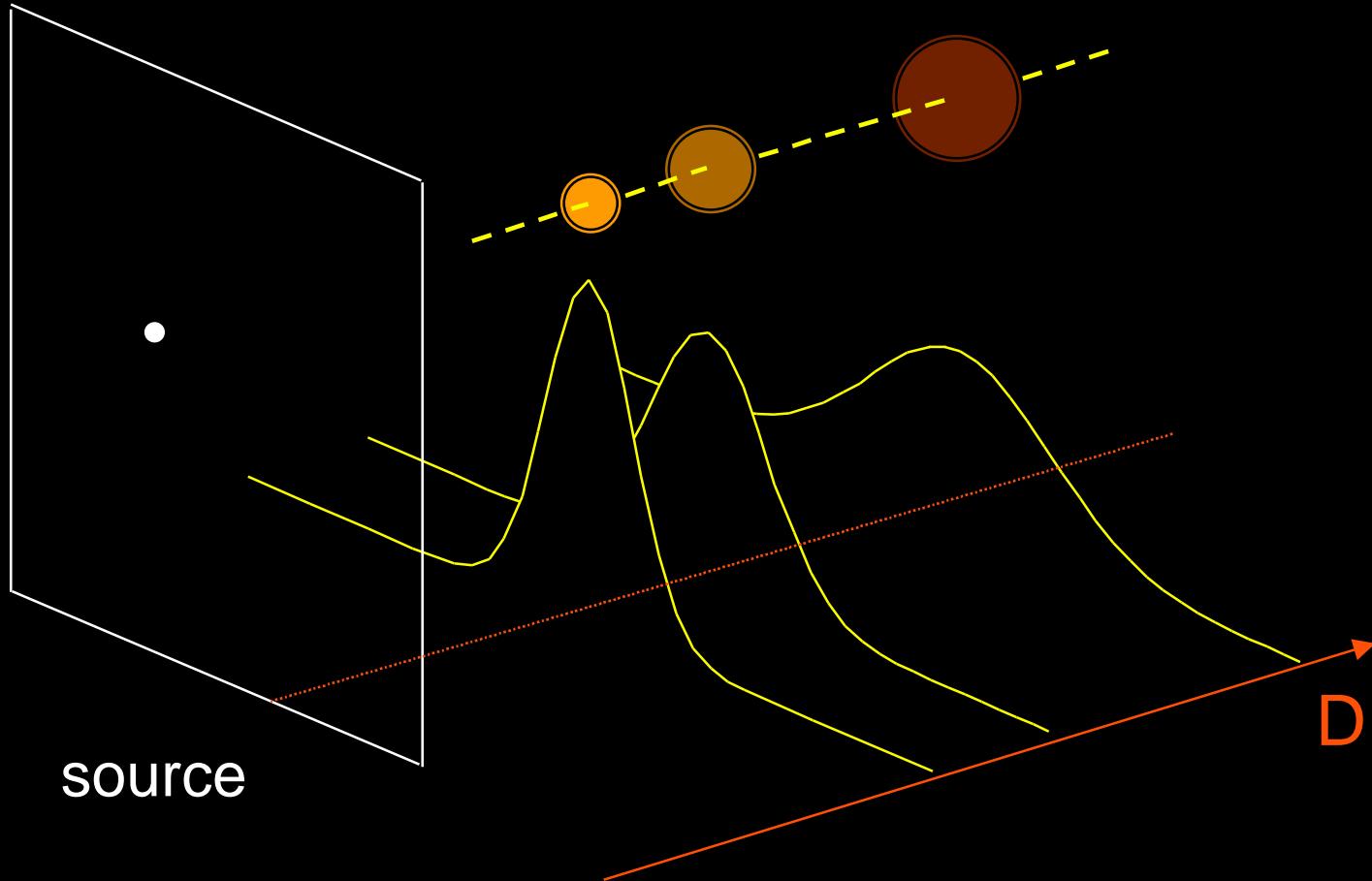


Correction de la diffusion en PET

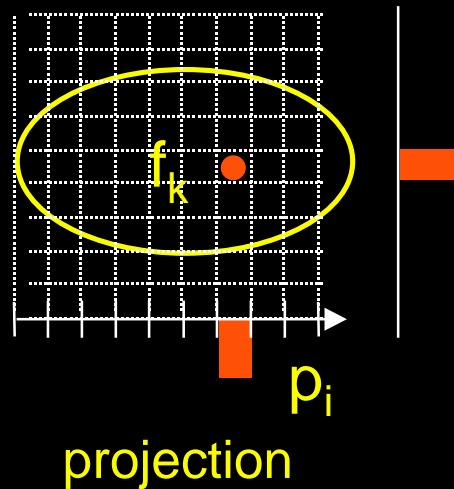
Ajustement du profil d'activité détecté en dehors du patient



Fonction de réponse du détecteur variable avec la profondeur



Correction de la fonction de réponse du détecteur en SPECT

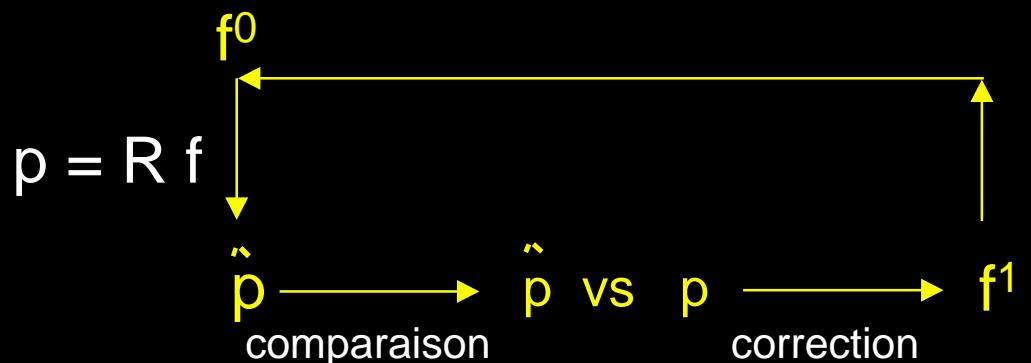


projection

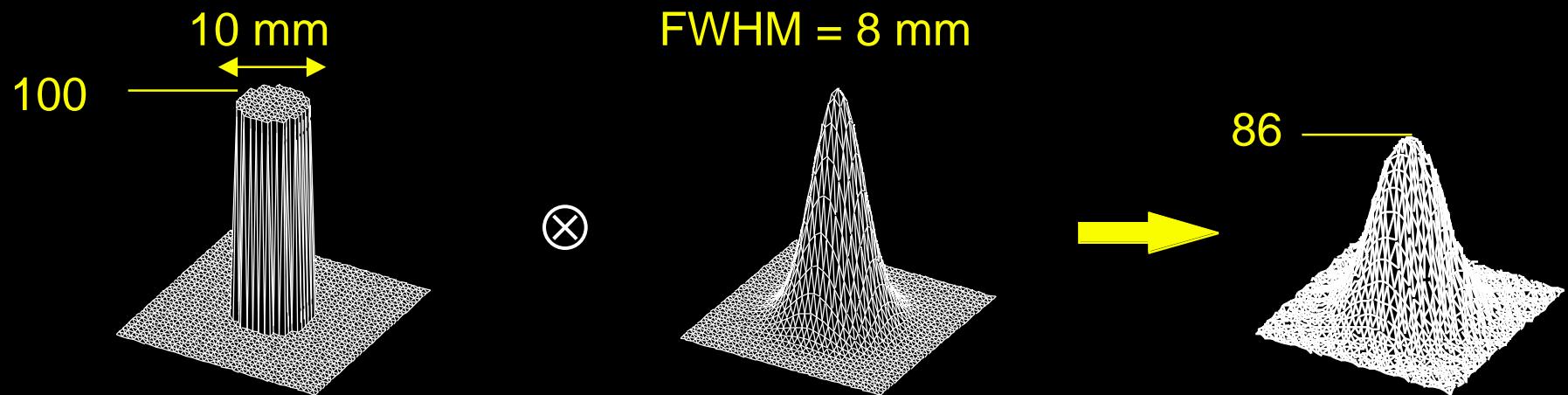
$$p = R f$$

$$\begin{bmatrix} p_1 \\ p_2 \\ p_3 \\ p_4 \end{bmatrix} = \begin{bmatrix} r_{11} & \dots & r_{14} \\ \vdots & \ddots & \vdots \\ r_{41} & \dots & r_{44} \end{bmatrix} \begin{bmatrix} f_1 \\ f_2 \\ f_3 \\ f_4 \end{bmatrix}$$

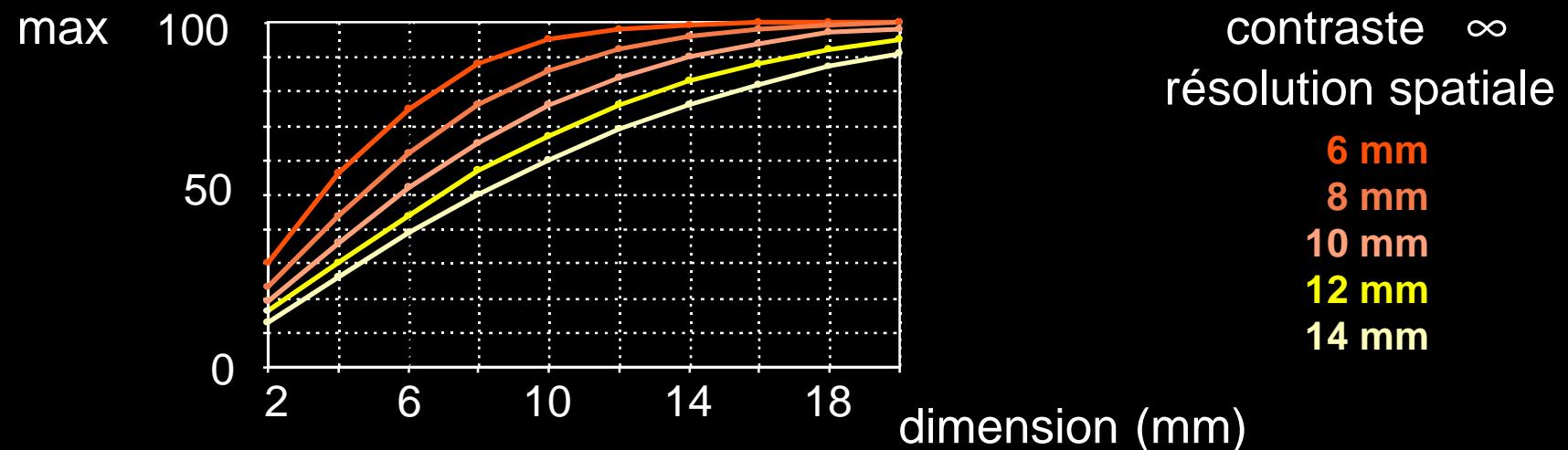
modélisation de la fonction de réponse du détecteur dans R



Effet de volume partiel

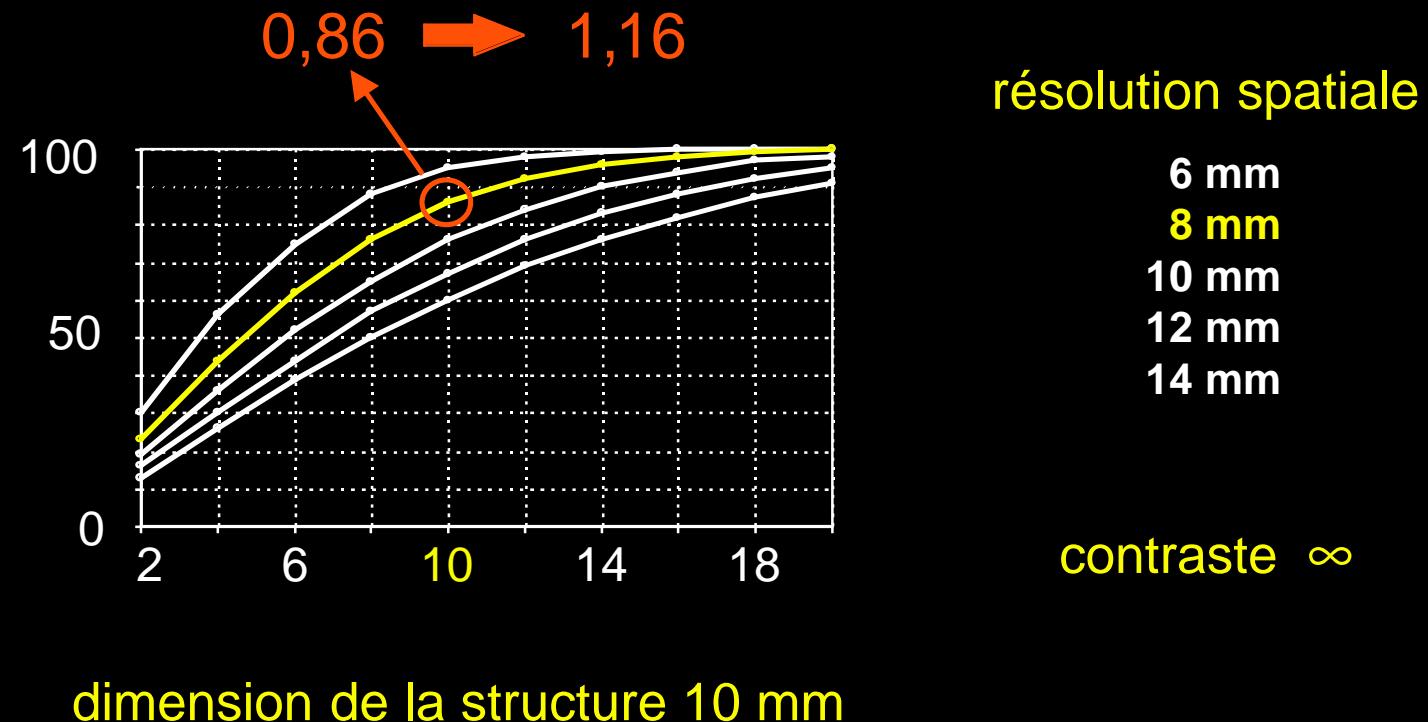


→ sous-estimation de la concentration réelle dans les structures de petite taille (typiquement < 2 FWHM)

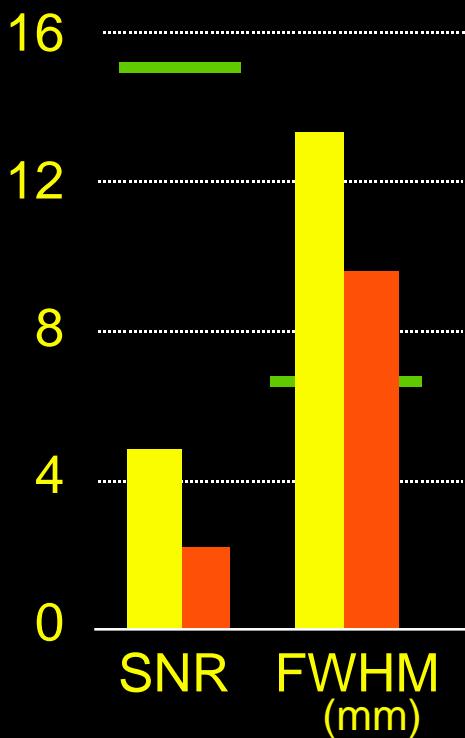
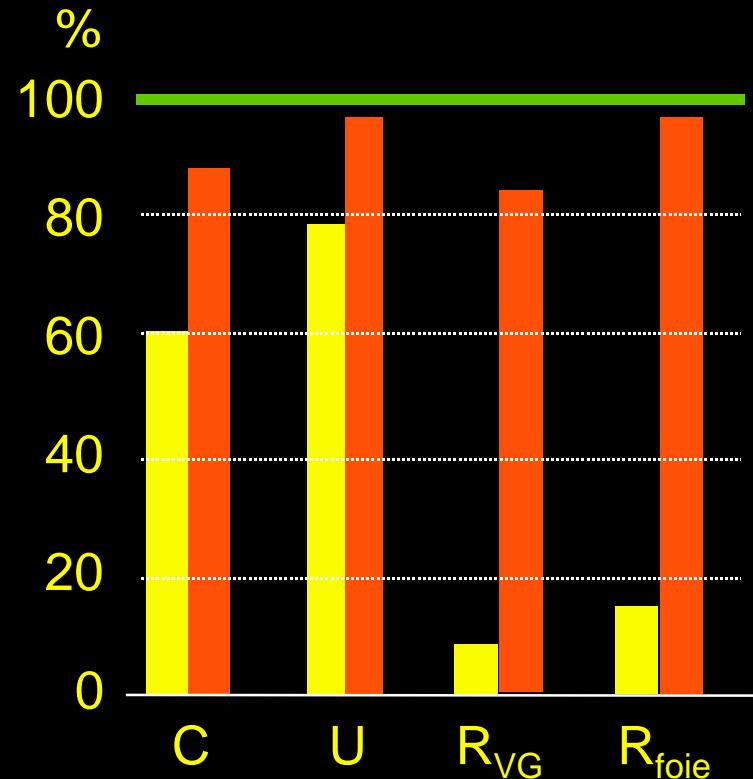


Solutions aux problèmes de volume partiel

Si taille des structures et contraste connus
→ coefficients de “recouvrement”



Importance des corrections en SPECT cardiaque au Tc-99m



sans correction
avec corrections
valeur optimale

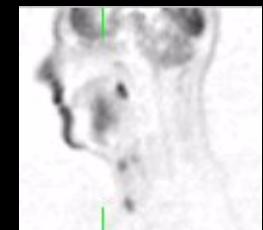
Caractéristiques des différents dispositifs

	SPECT	PET gamma-caméra	PET dédié
champ de vue axial (cm)	38	10	15
nb de coupes	95	15	35
résolution ds le plan (mm)	4,8	6,0	4,8
résolution axiale (mm)	5,5	6,0	4,0
sensibilité 2D (kcp/s)	-	-	220
sensibilité 3D (kcp/s)	100	91	1225

10,8 mCi FDG



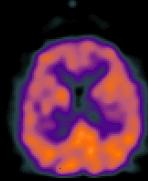
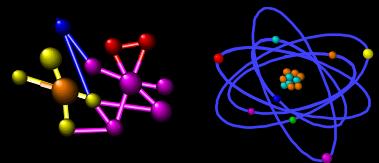
3h post inj.
DST-XL Orsay



4h post inj.
HR+

Futur

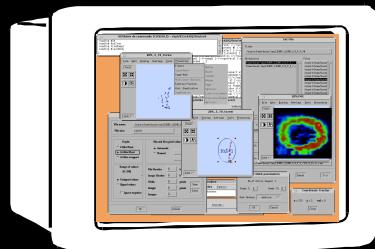
Radiotraceur



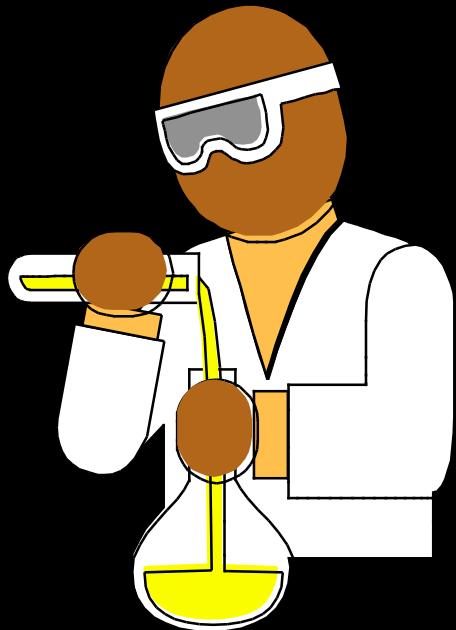
Détecteur



Traitement de l'information



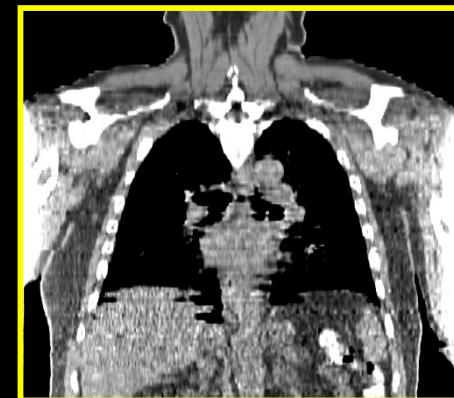
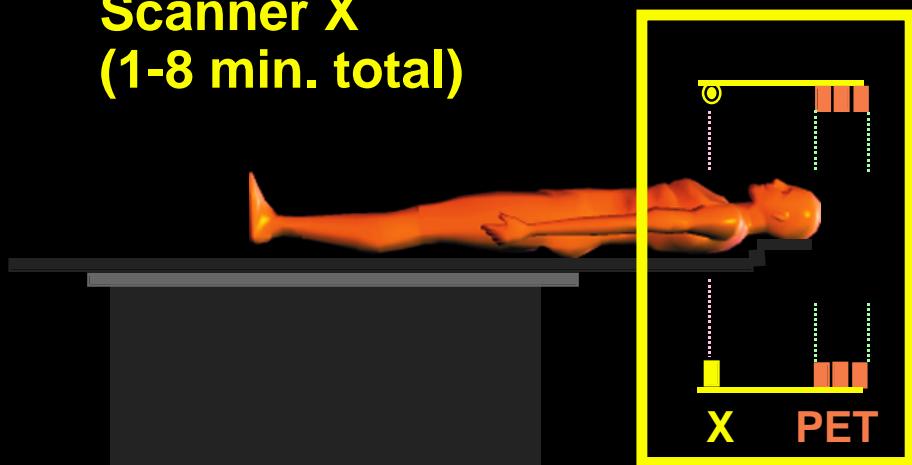
Nouveaux radiotraceurs



Oligonucléotides marqués pour la visualisation *in vivo* de l'expression des gènes...

Nouveaux détecteurs hybrides

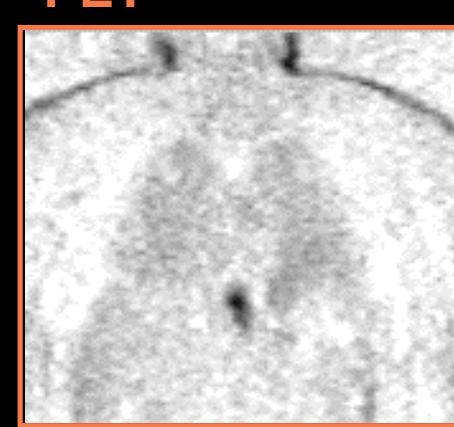
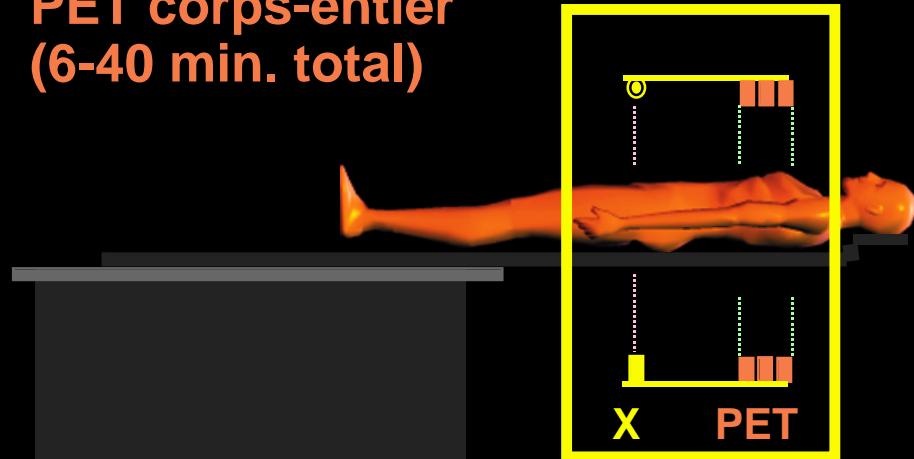
**Scanner X
(1-8 min. total)**



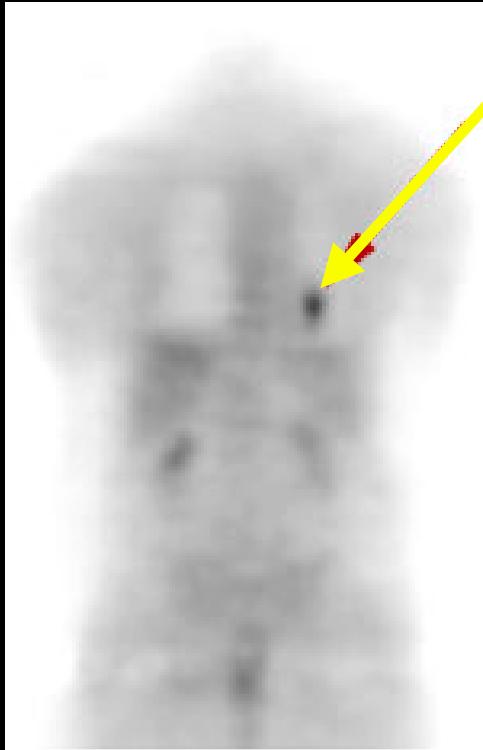
fusion



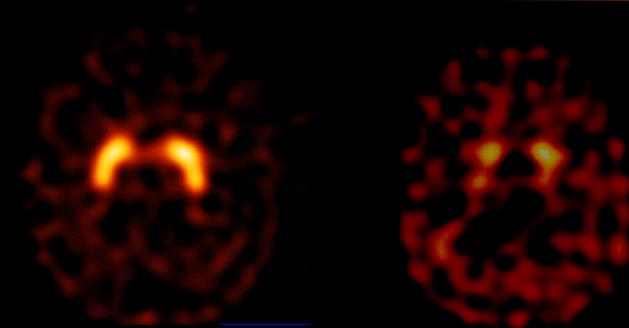
**PET corps-entier
(6-40 min. total)**



Quantification



X kBq/cc



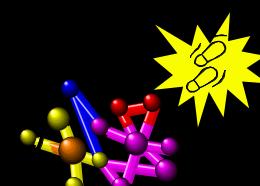
Diagnostic différentiel



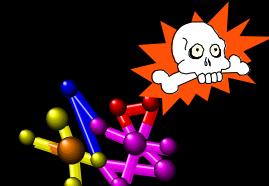
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