

TRAITEMENT & ANALYSE D'IMAGES MEDICALES

Application à l'évaluation pronostique des insuffisances cardiaques

Module IMAGE, Ecole doctorale I2S, Université Montpellier II

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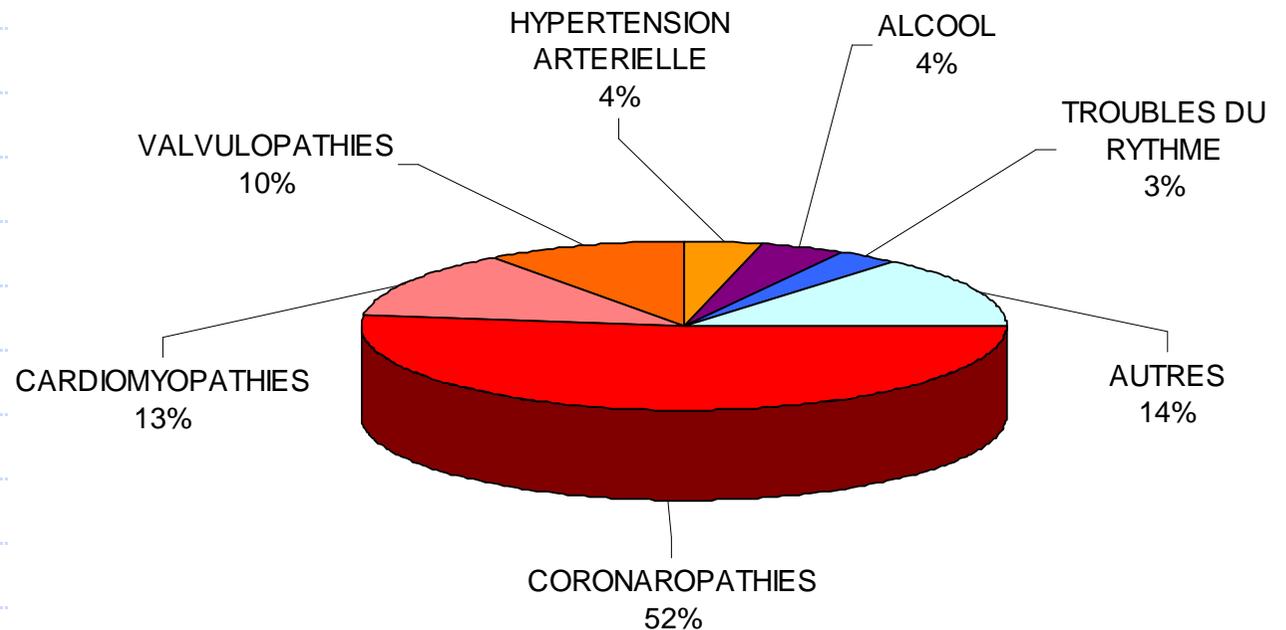
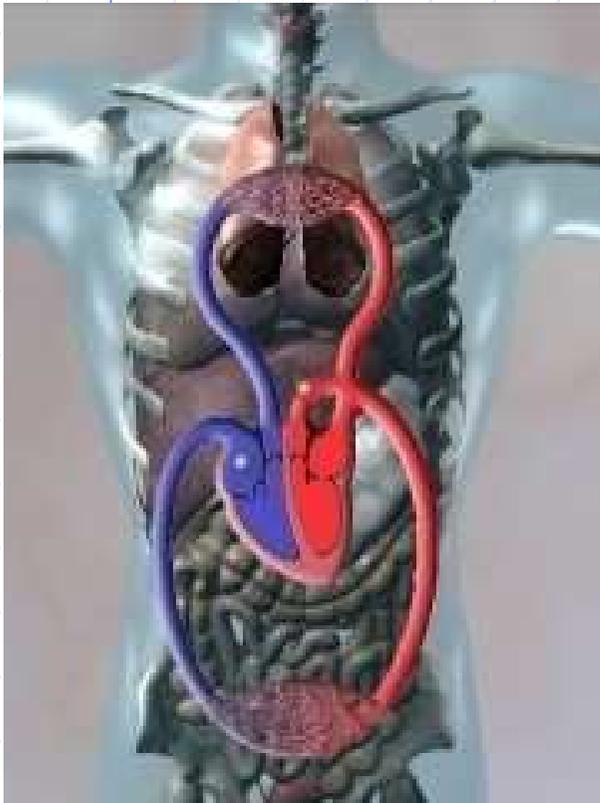
<http://scinti.etud.univ-montp1.fr>

PLAN

1. Problème médical à résoudre
2. Choix d'une modalité d'imagerie et d'un traceur
3. Acquisition des images
4. Reconstruction tomographique
5. Segmentation, analyse et validation (myocarde)
6. Segmentation, analyse et validation (ventriculo)

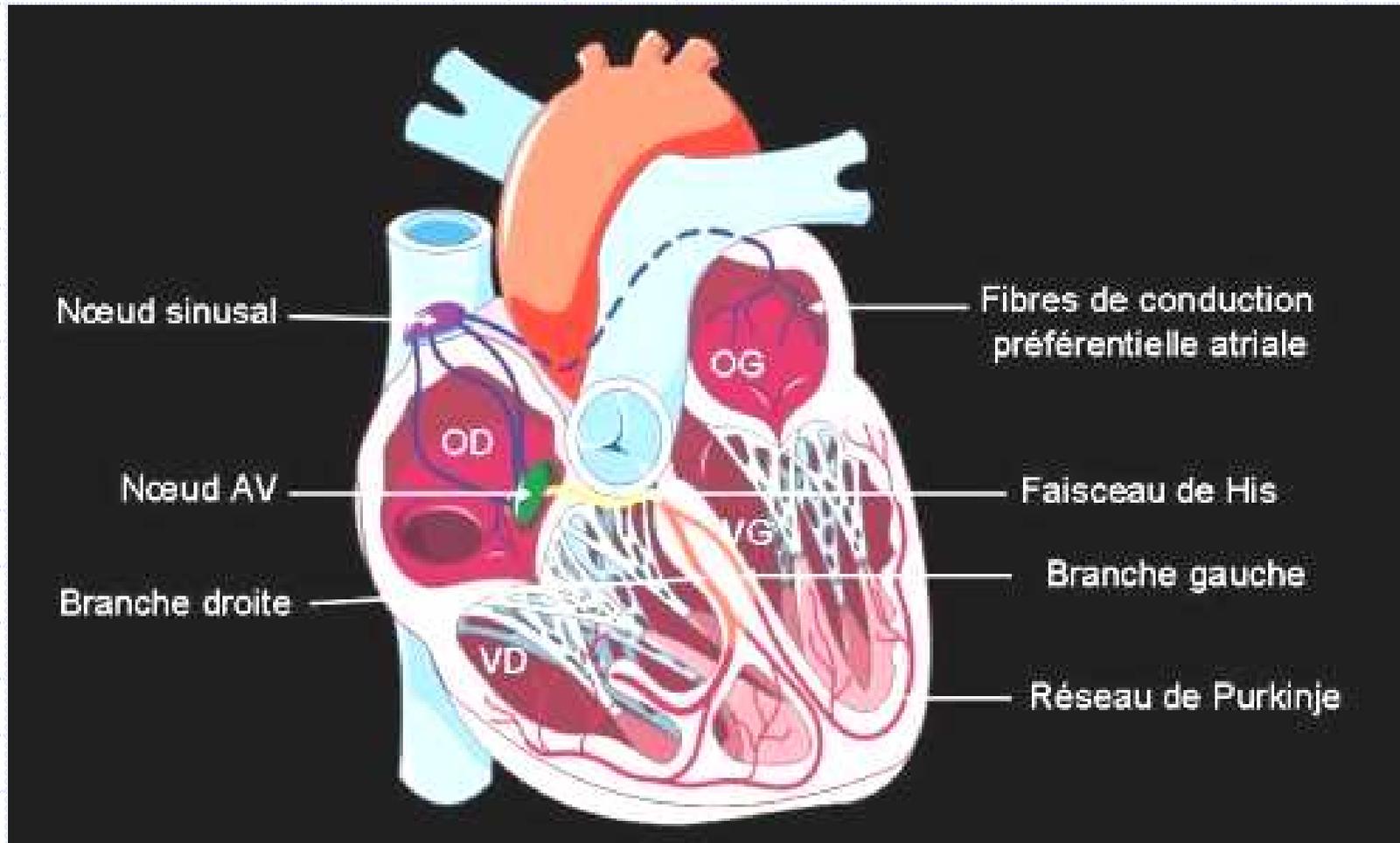
INSUFFISANCE CARDIAQUE

Insuffisance cardiaque : altération de la fonction pompe du coeur

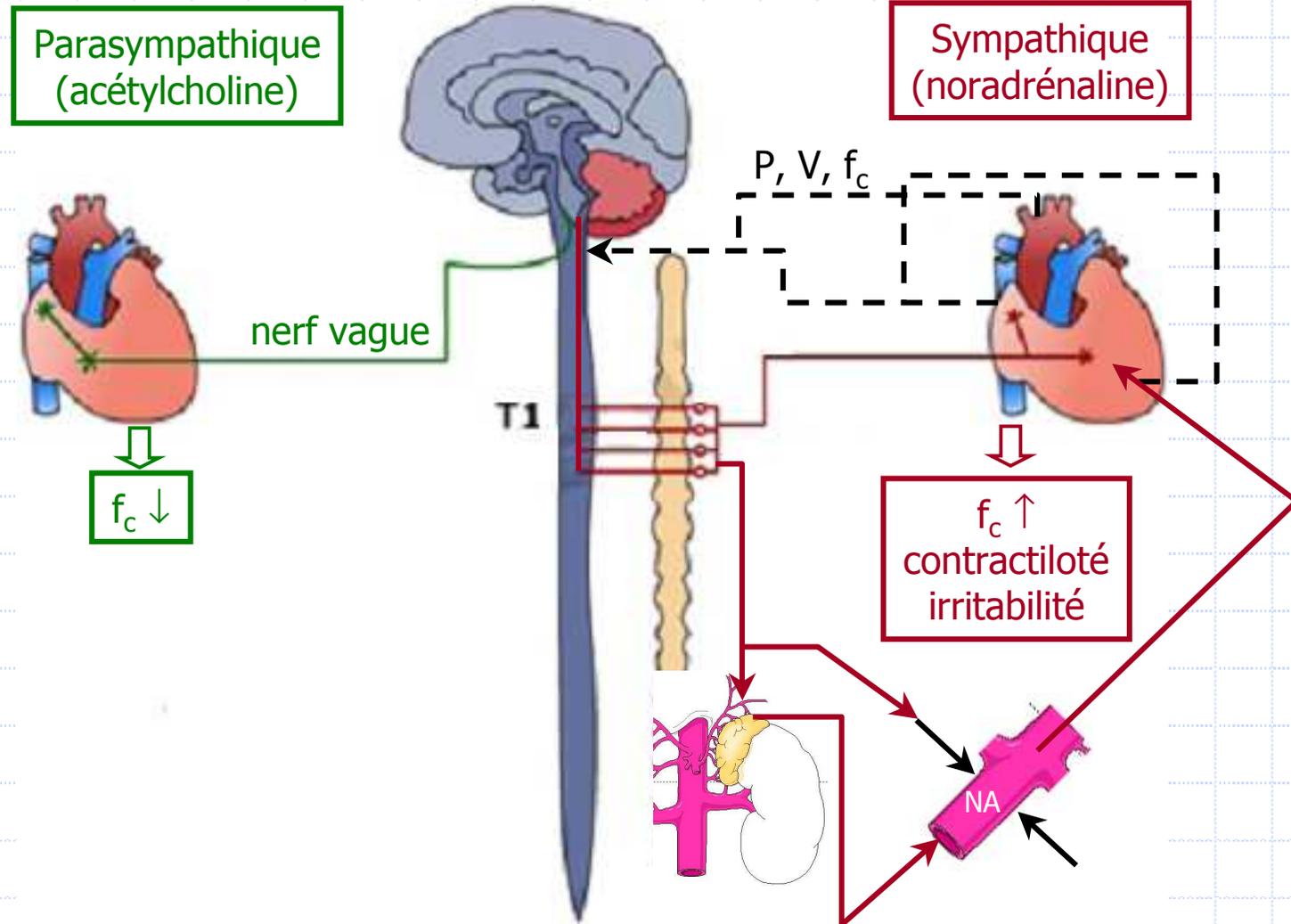


Prévalence = 1-3 % (1 million en France)
A 5 ans : 75 (H) à 62%(F) de survie
60% à un an si stade IV
Environ 1/2 morts subites

LA POMPE CARDIAQUE

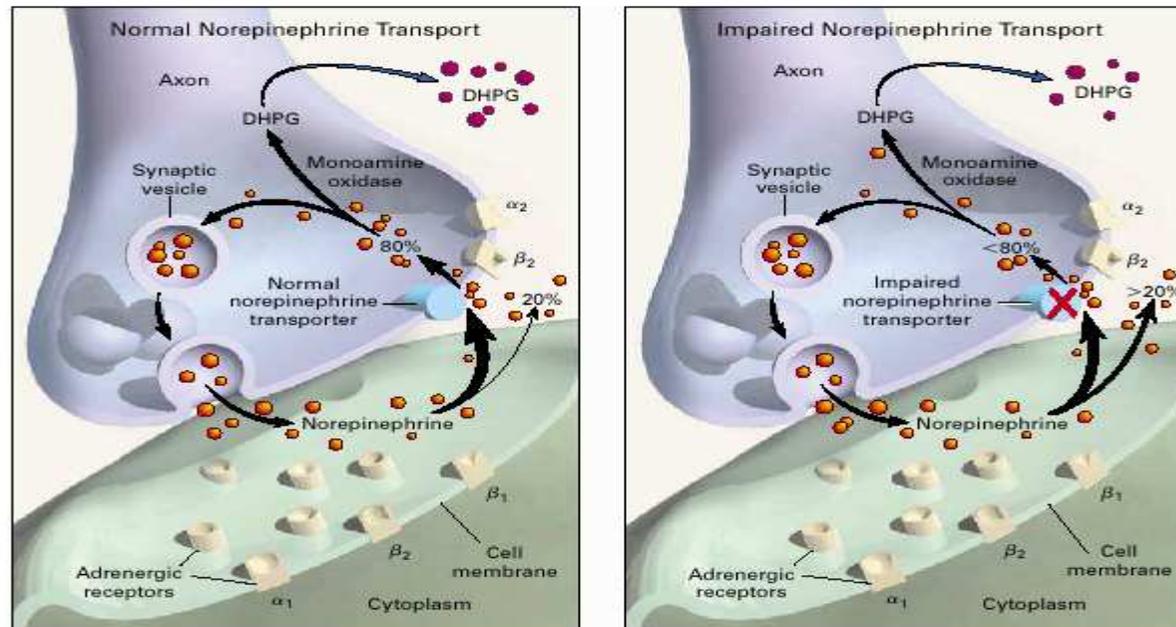


REGULATION



CAUSES DES DECES

- ◆ Troubles du rythme, embolies,...
- ◆ Augmentation de l'activité sympathique
 - ◆ Remodelage, ↓ fonction pompe, ↑ symptômes
 - ◆ ↓ recapture de la NA: désensibilisation, courts-circuits



TRAITEMENTS

Règles hygiéno-diététiques

- Activité physique
- Régime hyposodé

Traitements médicamenteux

- Diurétiques
- Inhibiteur enzyme conversion
- Bêta-bloquant, anti-aldostérone

TRAITEMENTS

Règles hygiéno-diététiques

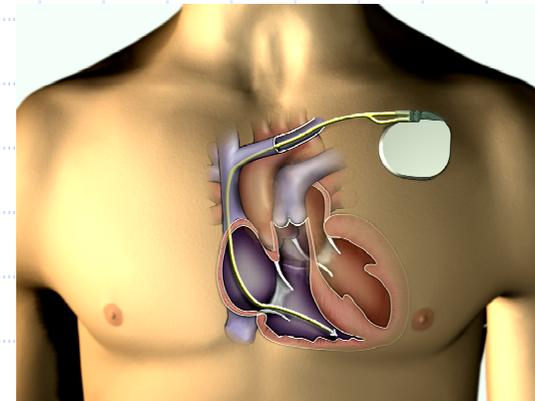
- Activité physique
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Traitements médicamenteux

- Diurétiques
- Inhibiteur enzyme conversion
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Traitements électriques

- Resynchronisation
- **Défibrillateur**



Assistance

Greffe cardiaque



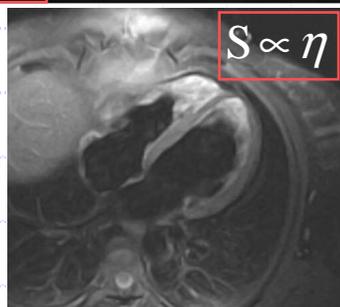
PROBLEME CLINIQUE

- ◆ Indication de défibrillateur implantable sur la
 $FE = (VTD-VTS)/VTD < 35\% \text{ ou } 40\%$
 - ◆ 35 % seulement des défibrillateurs implantés sont sollicités sur un suivi de 3 ans.
 - ◆ Des décès surviennent avec $FE > 40\%$

- ◆ Identification des patients à risque de mort subite par imagerie ?
 - ◆ Quelle modalité d'imagerie ?
 - Mouvements myocardiques, fonction pompe ?
 - innervation sympathique ?
 - ◆ Quels traitements d'image ?

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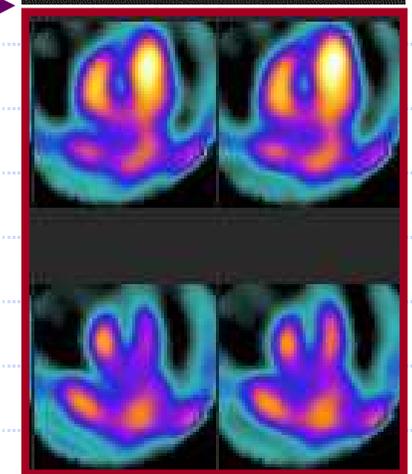
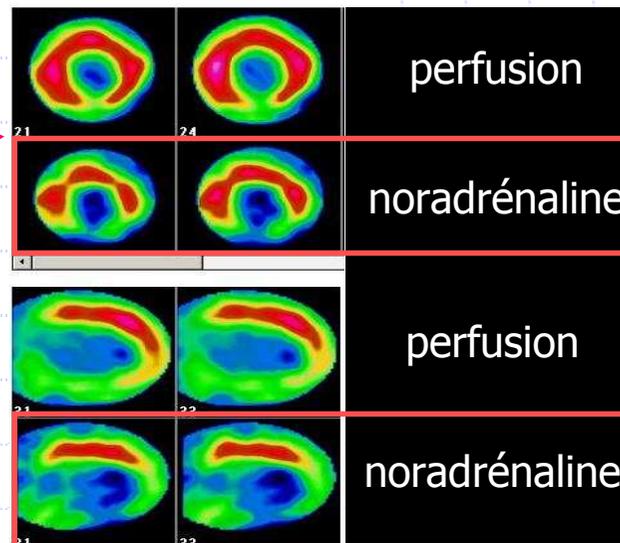
CHOIX D'UNE MODALITE D'IMAGERIE



ANATOMIQUE

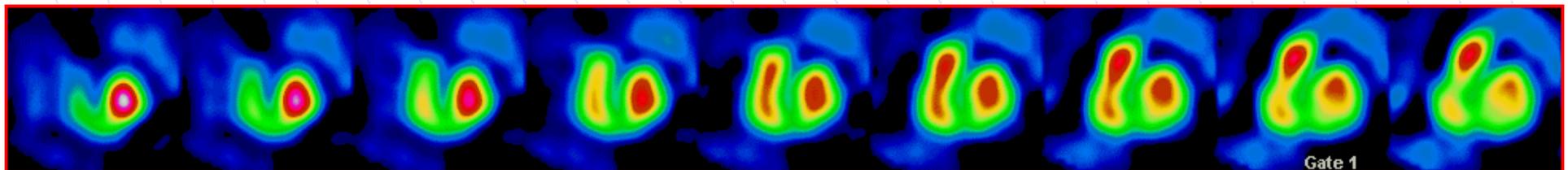
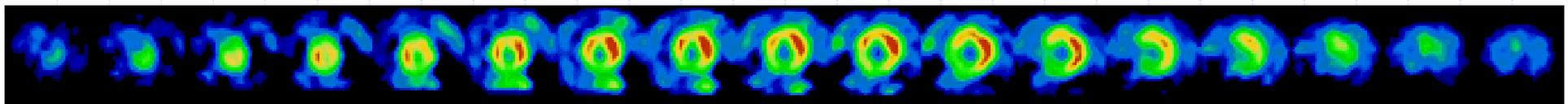
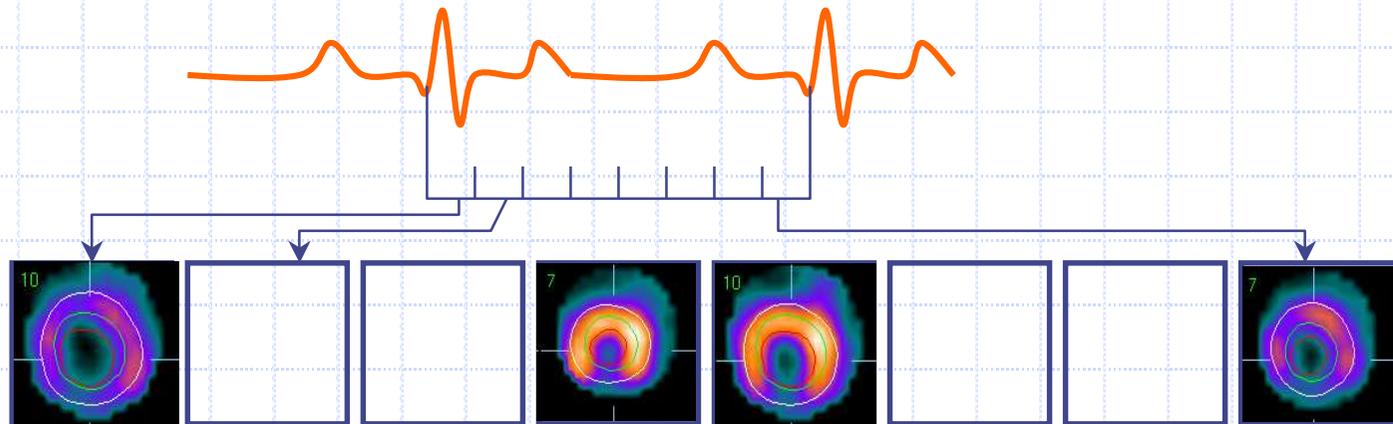
METABOLIQUE

FONCTIONNELLE



$C = f(t)$

SYNCHRONISATION A L'ECG



CHOIX D'UN TRACEUR RADIOACTIF

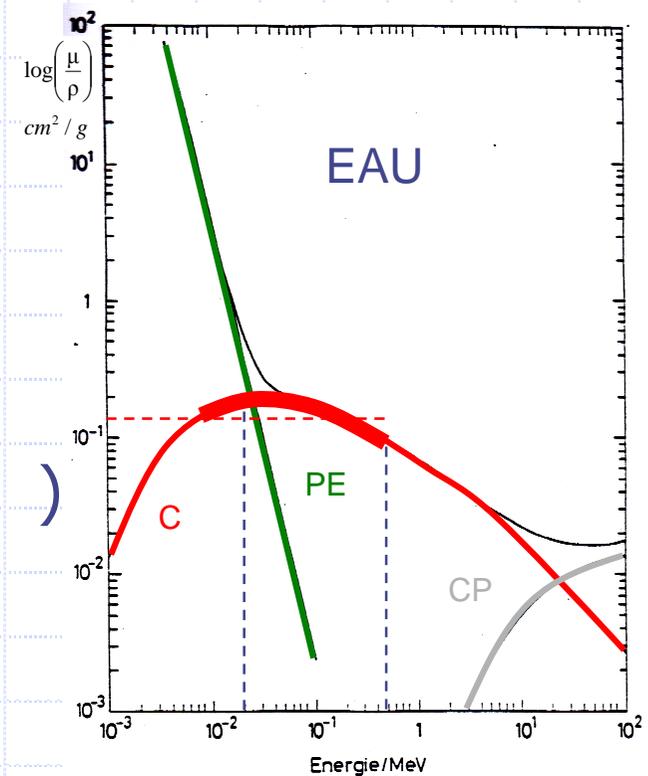
◆ Photons pénétrants et peu diffusés

- ◆ Émetteurs gamma ou bêta plus (ionisants)

$$\mu_{PE} \propto \frac{\rho \cdot Z^3}{E_{\varphi}^3} \quad \mu_C \propto \rho$$

◆ énergies détectables

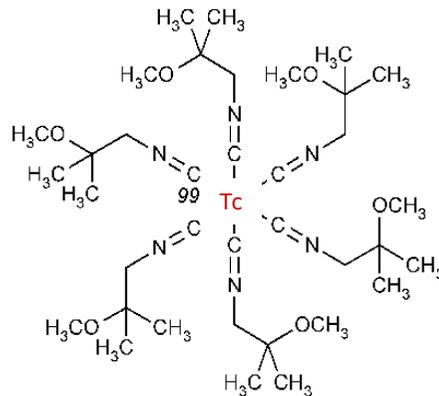
- ◆ γ (140 keV $^{99}_{43}\text{Tc}$ ou 159 keV $^{123}_{53}\text{I}$)
- ◆ $\beta^+ \rightarrow 2\gamma$ 511 keV ($^{18}_8\text{F}$)



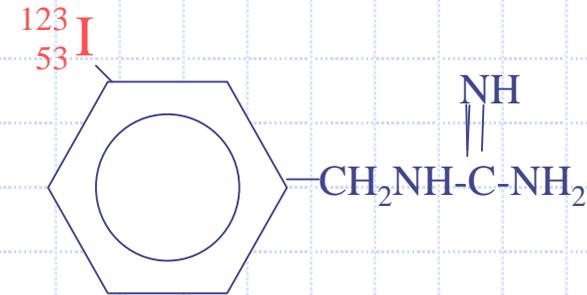
CHOIX D'UN TRACEUR RADIOACTIF

◆ Traçage possible d'un métabolisme

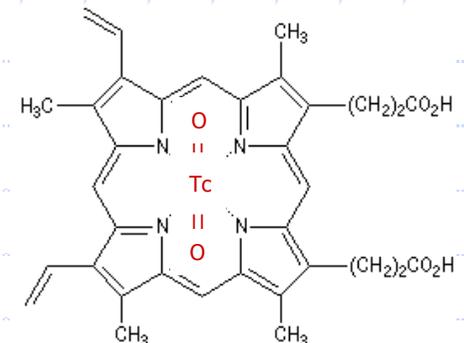
- ◆ Marquage de **cations lipophiles (MIBI)** au $^{99}_{43}\text{Tc}$



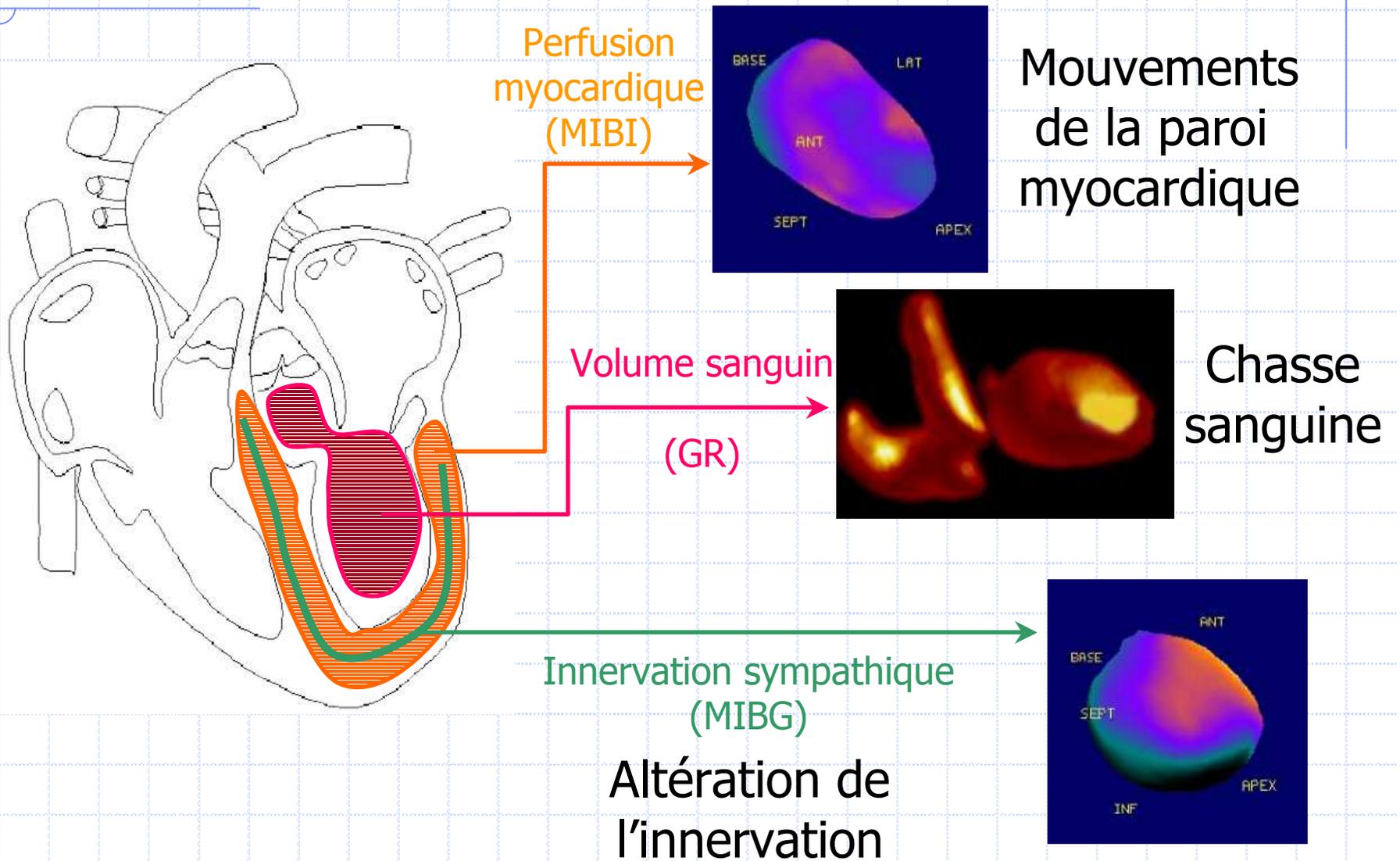
- ◆ Marquage de la **MIBG** à $^{123}_{53}\text{I}$



- ◆ Marquage des **globules rouges** au $^{99}_{43}\text{Tc}$
 - Chimie de complexation avec le noyau tétrapyrrolique de l'hème

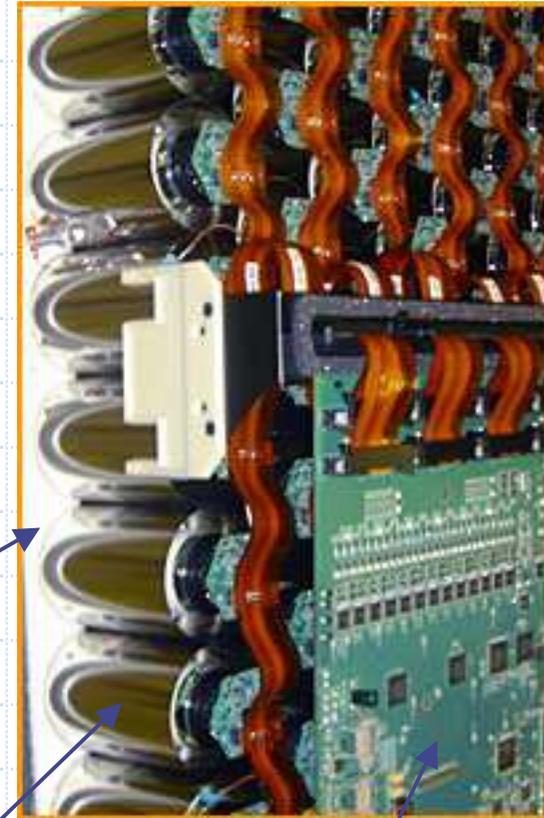


CHOIX D'UNE MODALITE D'IMAGERIE



1. Problème médical à résoudre
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Single Photon Emission CT



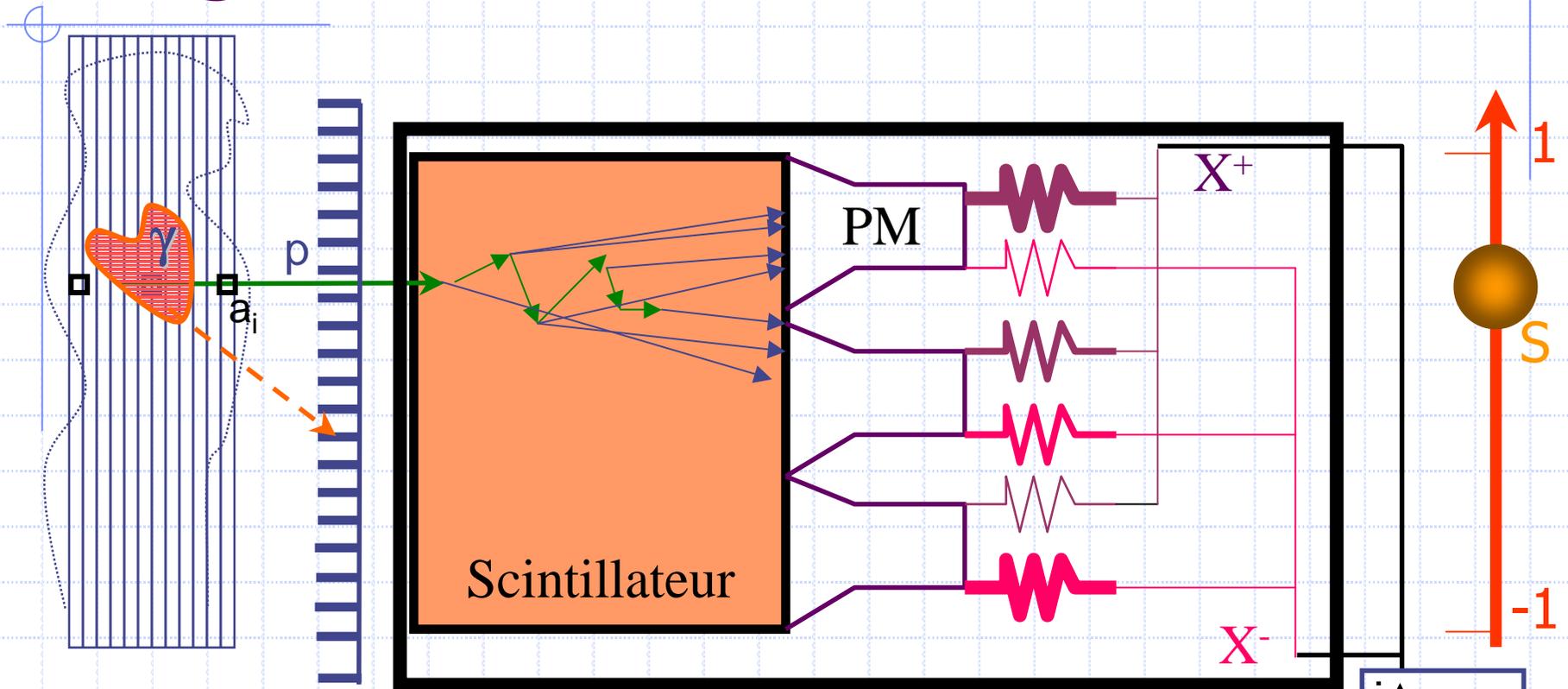
Collimateur

Scintillateur

Photo-multiplicateur

Localisation

Single Photon Emission CT

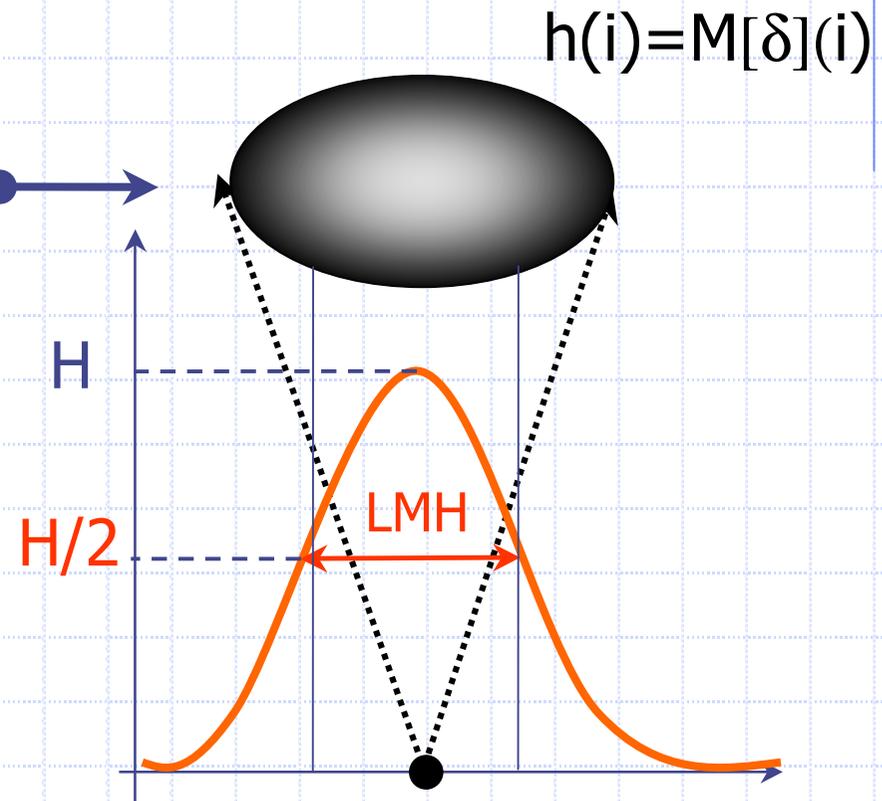
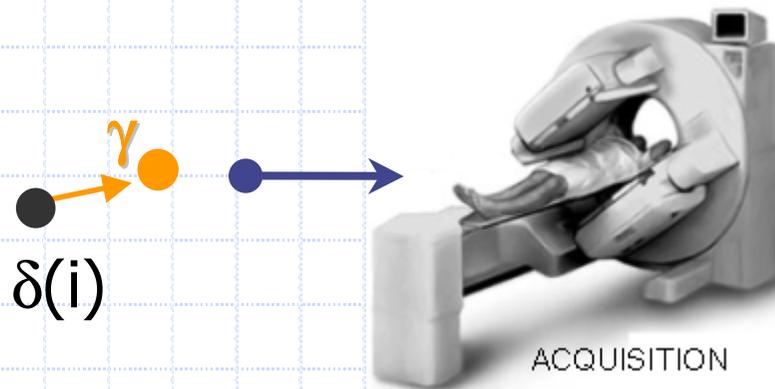


Spectroscopie

Localisation

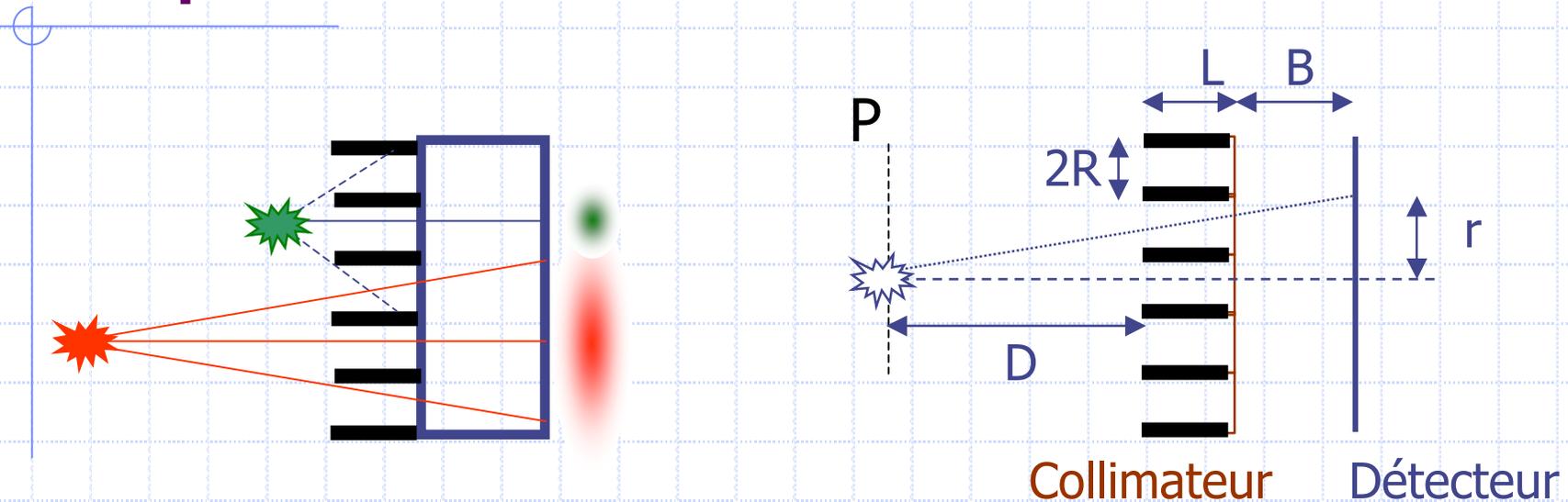
$$X = \frac{X^+ - X^-}{X^+ + X^-}$$

Largeur à mi-hauteur (FWHM)



- Réponse intrinsèque
 - ◆ $LMH \leq 4 \text{ mm}$, \pm invariante
- Réponse du collimateur
 - ◆ La LMH dépend de la distance source-détecteur

Réponse d'un collimateur

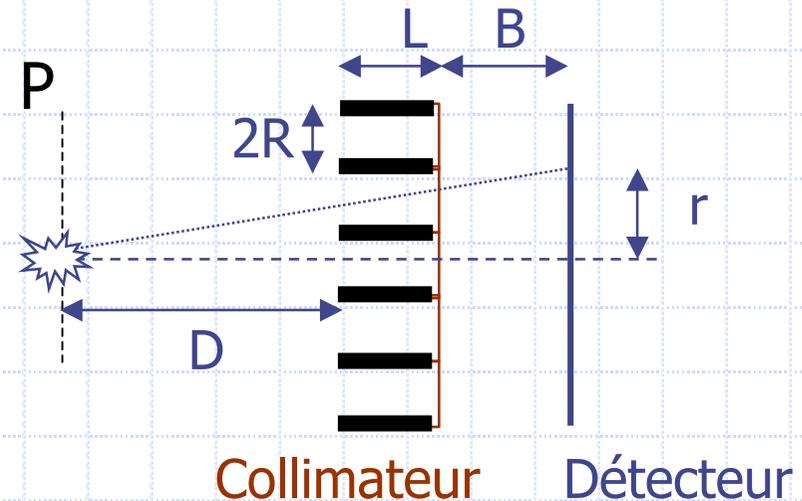
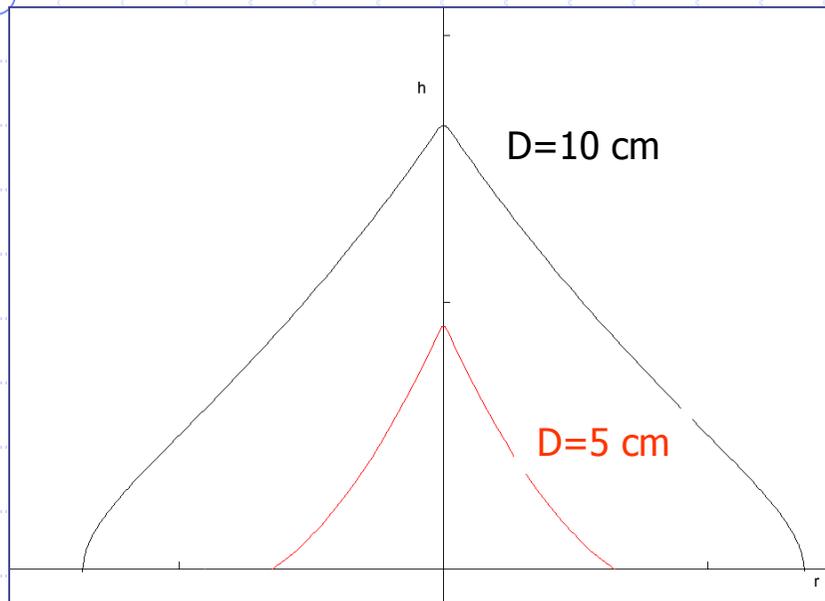


Réponse moyenne dans le plan P (septa cylindriques) :

$$h(r) = \frac{\varepsilon}{\pi} \left[2 \cdot \arccos\left(\frac{\lambda \cdot r}{2}\right) - \lambda \cdot r \cdot \sqrt{1 - \left(\frac{\lambda \cdot r}{2}\right)^2} \right] \quad \lambda = \frac{L}{R \cdot (L + D + B)}$$

ε : efficacité du collimateur

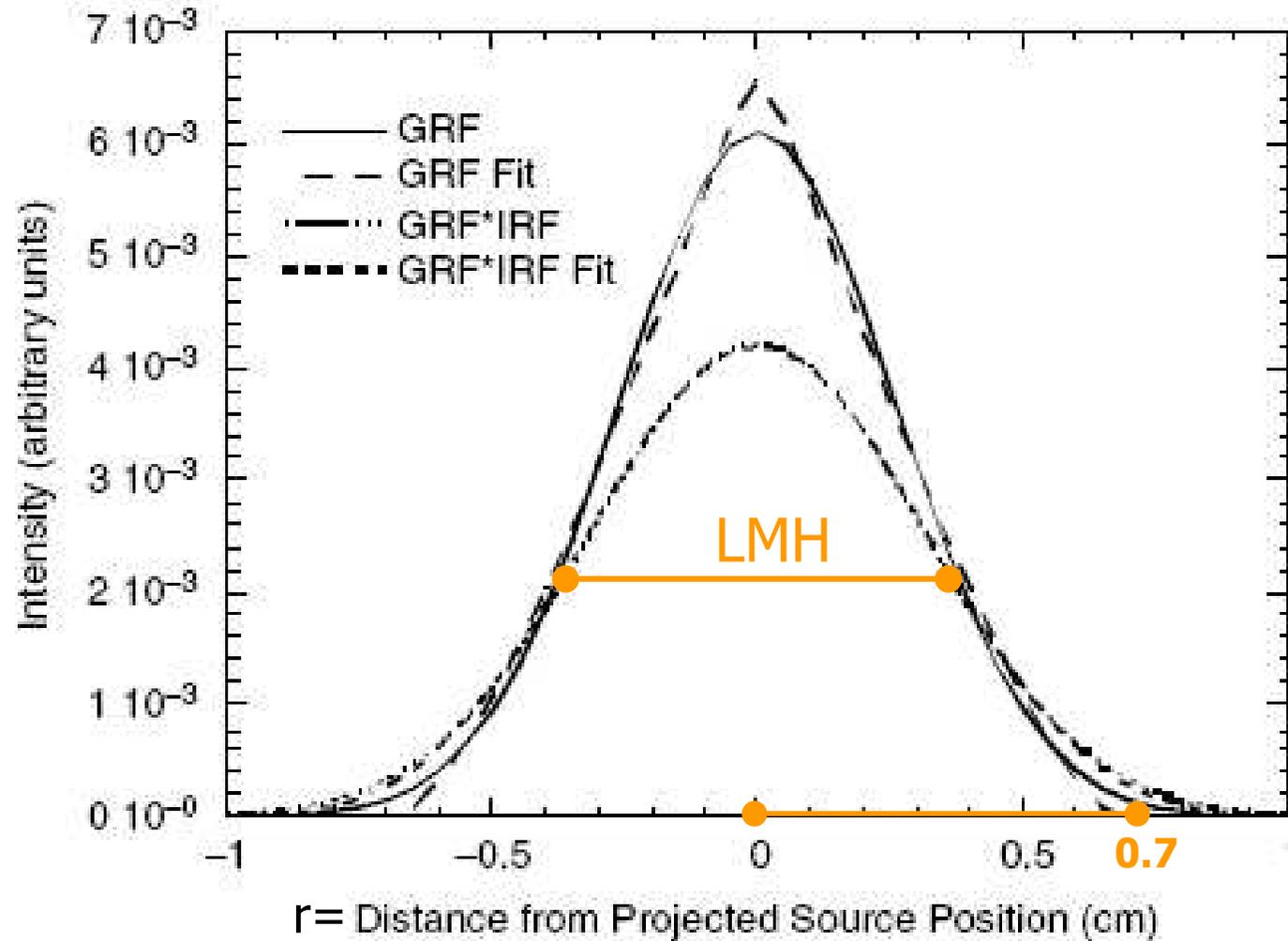
Réponse d'un collimateur



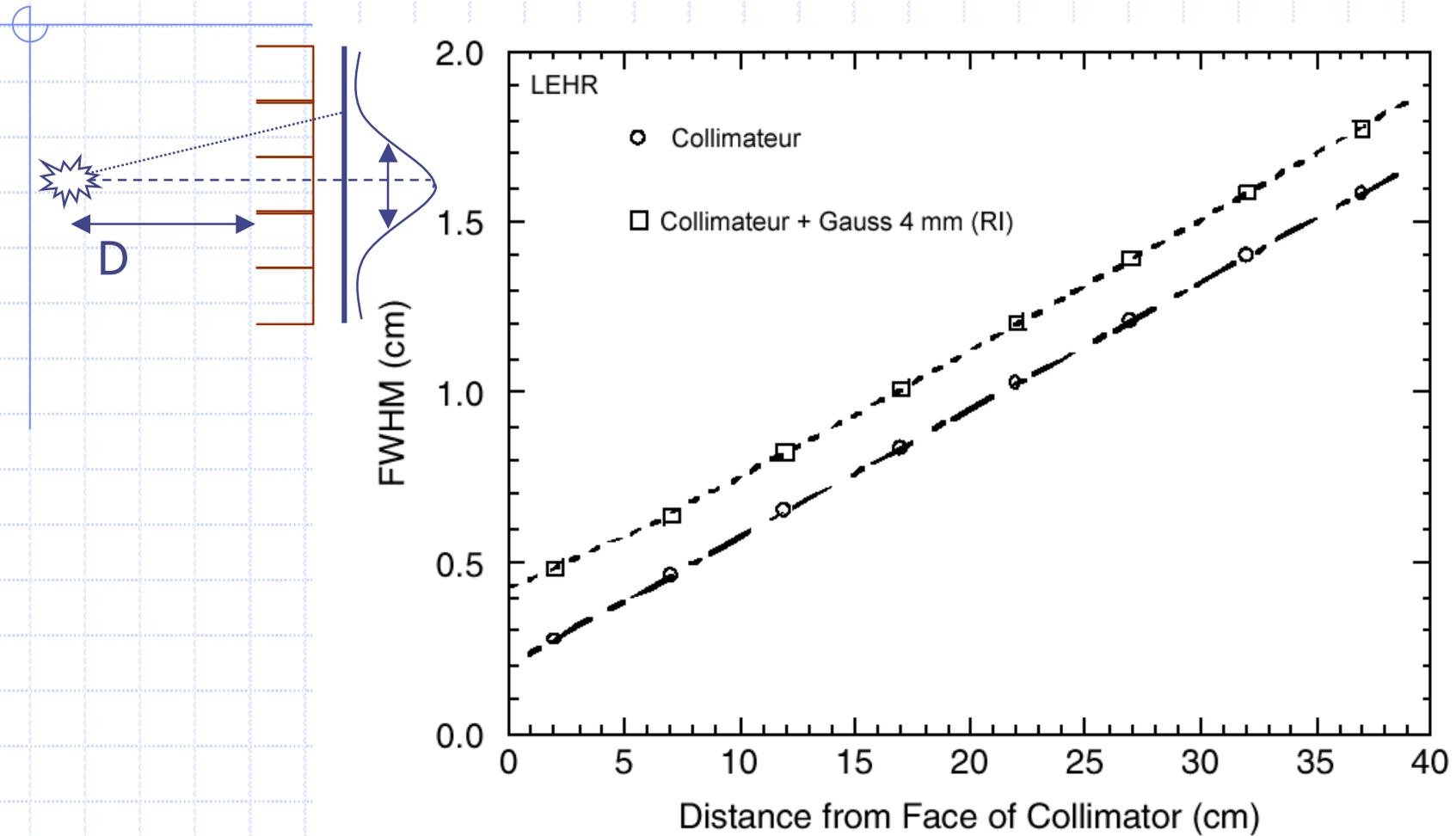
$$h(r) = \frac{\varepsilon}{\pi} \left[2 \cdot \arccos\left(\frac{\lambda \cdot r}{2}\right) - \lambda \cdot r \cdot \sqrt{1 - \left(\frac{\lambda \cdot r}{2}\right)^2} \right] \quad \lambda = \frac{L}{R \cdot (L + D + B)}$$

LEHR : L = 4,1 cm ; B = 0,64 cm ; R = 0,19 cm ; $\varepsilon = 0,065$

Réponse d'une γ -caméra \approx gaussienne



Linéarité entre LMH et D

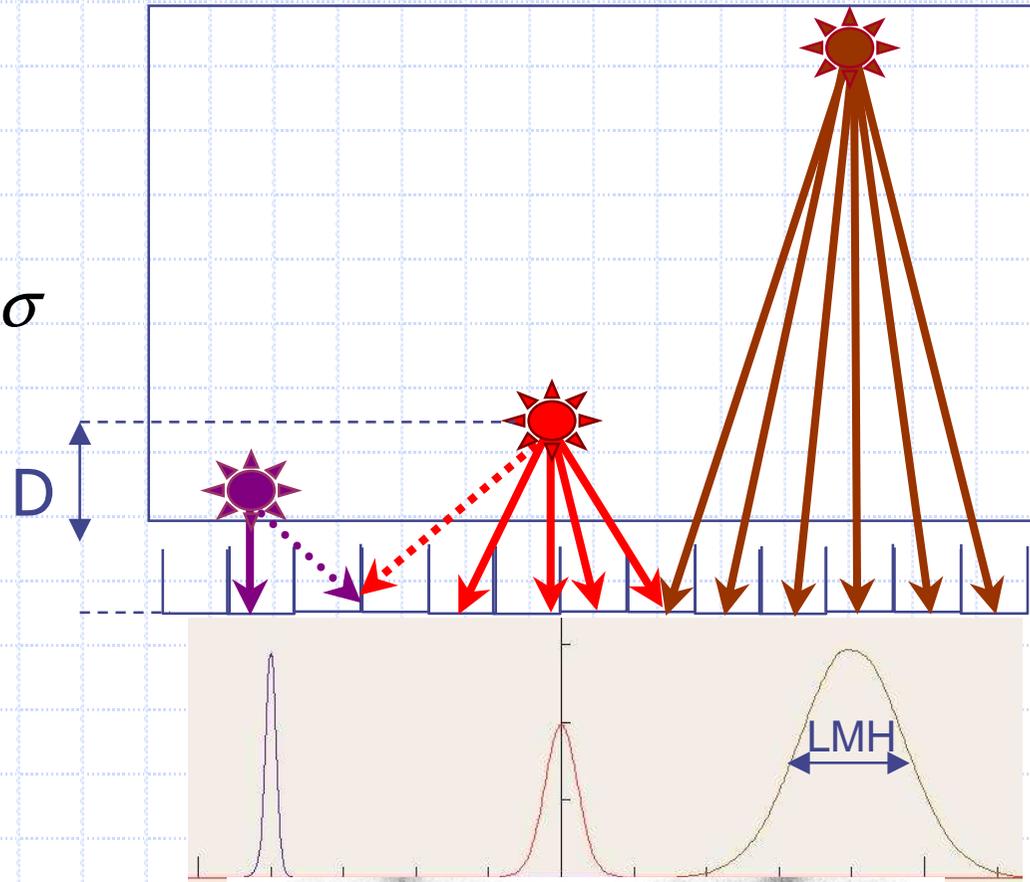
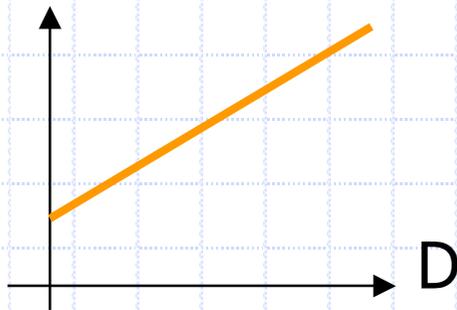


Réponse d'une γ -caméra

$$h(i) = \frac{1}{\sqrt{2\pi}\sigma} e^{-\frac{i^2}{2\sigma^2}}$$

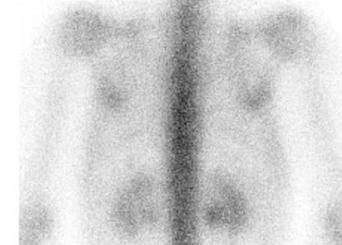
$$LMH = 2 \cdot \sqrt{2 \cdot \ln 2} \sigma \approx 2,35 \cdot \sigma$$

$$\sigma = k \cdot D + k'$$



Utilité d'une déconvolution
(non stationnaire)

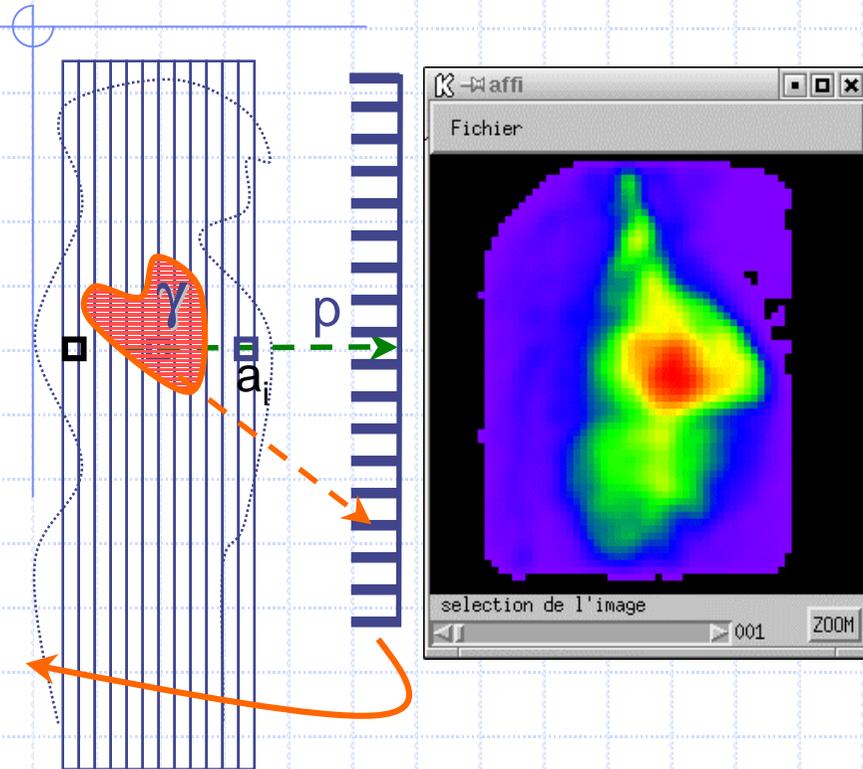
contact



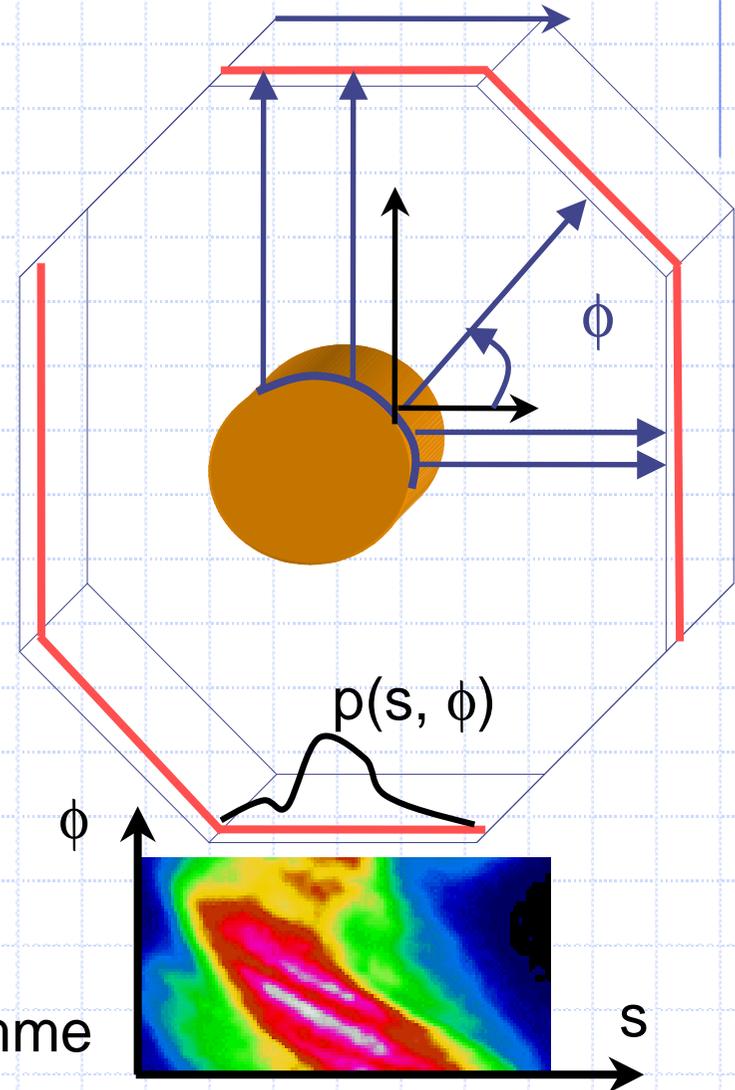
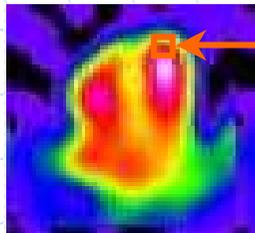
50 cm

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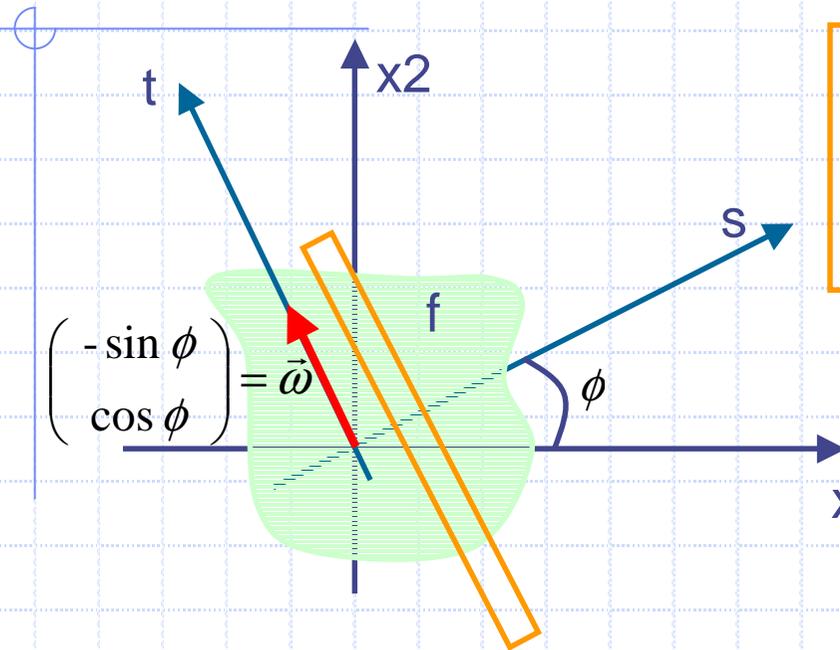


$$p = a_1 + a_2 + \dots + a_n$$



sinogramme

Modélisation analytique

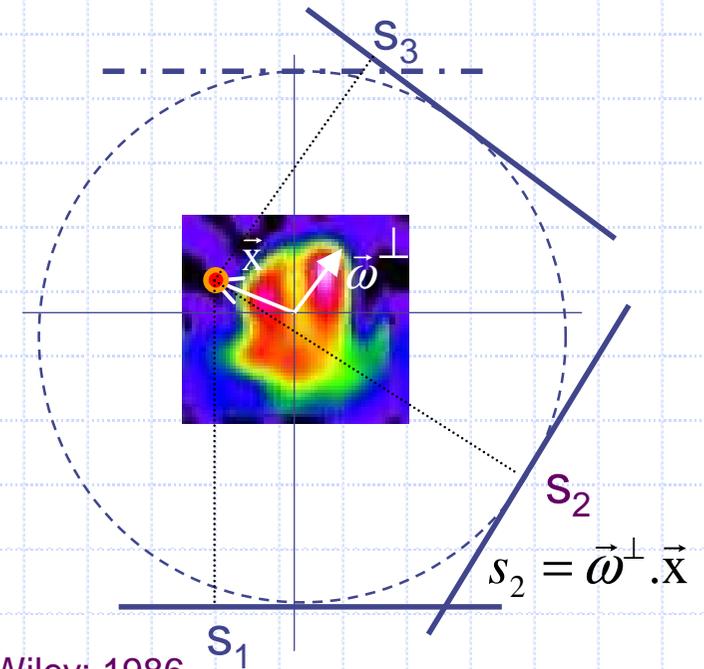


$$p(\vec{\omega}, s) = p_{\vec{\omega}}(s) = \int_t f(s \vec{\omega}^\perp + t \vec{\omega}) dt$$

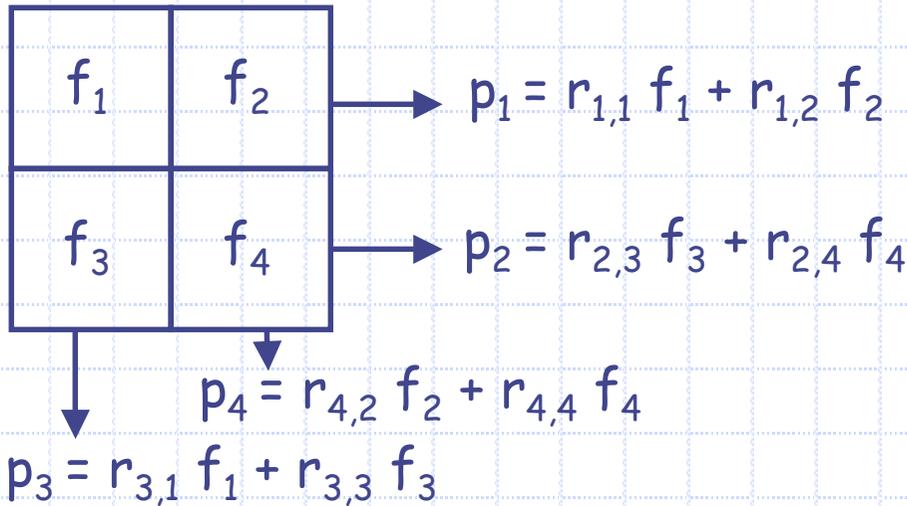
$p = Rf$ transformée de Radon

$$(R^* p)(\vec{x}) = \int_{\phi=0}^{\pi} p(\vec{\omega}, \vec{\omega}^\perp \cdot \vec{x}) d\phi$$

rétroprojection = épandage



Modélisation algébrique



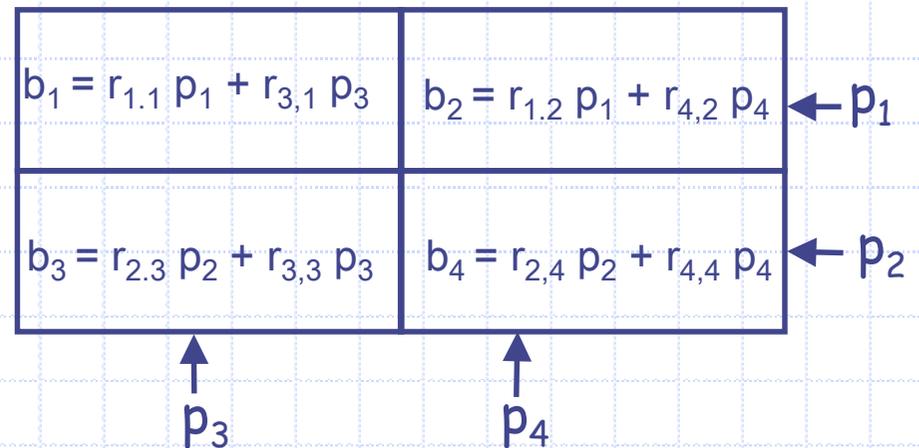
$$\begin{pmatrix} r_{1,1} & r_{1,2} & r_{1,3} & r_{1,4} \\ r_{2,1} & r_{2,2} & r_{2,3} & r_{2,4} \\ r_{3,1} & r_{3,2} & r_{3,3} & r_{3,4} \\ r_{4,1} & r_{4,2} & r_{4,3} & r_{4,4} \end{pmatrix} \cdot \begin{pmatrix} f_1 \\ f_2 \\ f_3 \\ f_4 \end{pmatrix} = \begin{pmatrix} p_1 \\ p_2 \\ p_3 \\ p_4 \end{pmatrix}$$

$r_{i,j}$ = % du pixel j intersecté par la projection i

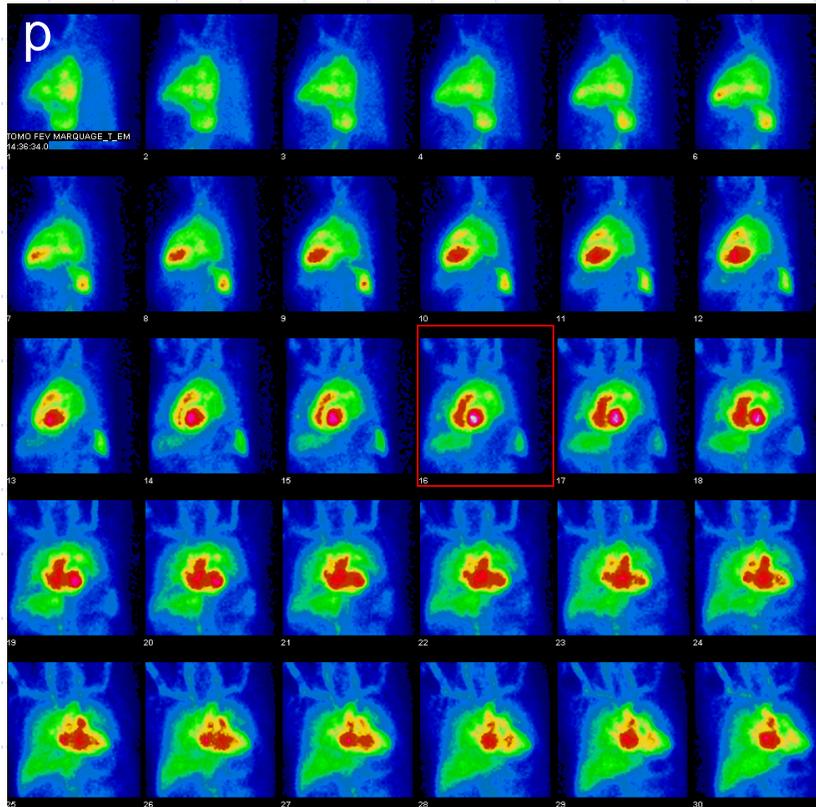
$$\mathbf{R} \cdot \vec{f} = \vec{p}$$

$$\begin{pmatrix} r_{1,1} & r_{2,1} & r_{3,1} & r_{4,1} \\ r_{1,2} & r_{2,2} & r_{3,2} & r_{4,2} \\ r_{1,3} & r_{2,3} & r_{3,3} & r_{4,3} \\ r_{1,4} & r_{2,4} & r_{3,4} & r_{4,4} \end{pmatrix} \cdot \begin{pmatrix} p_1 \\ p_2 \\ p_3 \\ p_4 \end{pmatrix} = \begin{pmatrix} b_1 \\ b_2 \\ b_3 \\ b_4 \end{pmatrix}$$

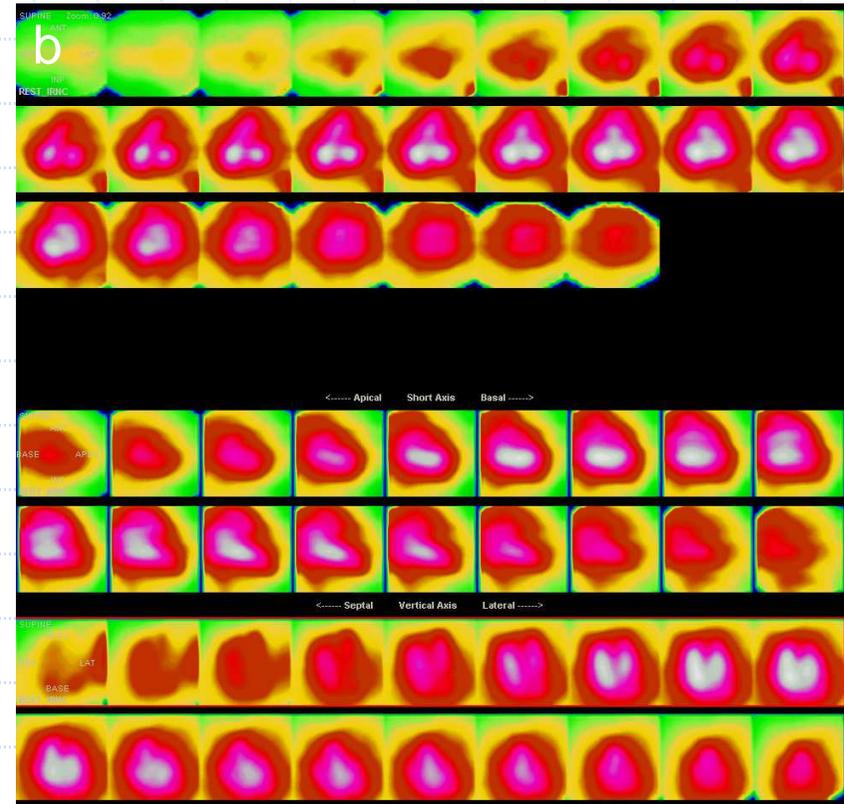
$${}^t\mathbf{R} \cdot \vec{p} = \vec{b}$$



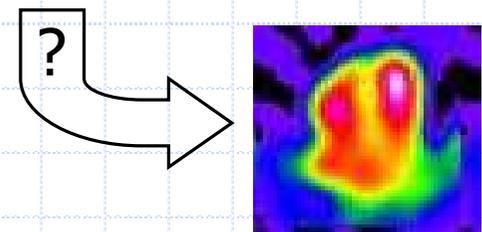
Projection / Rétroprojection



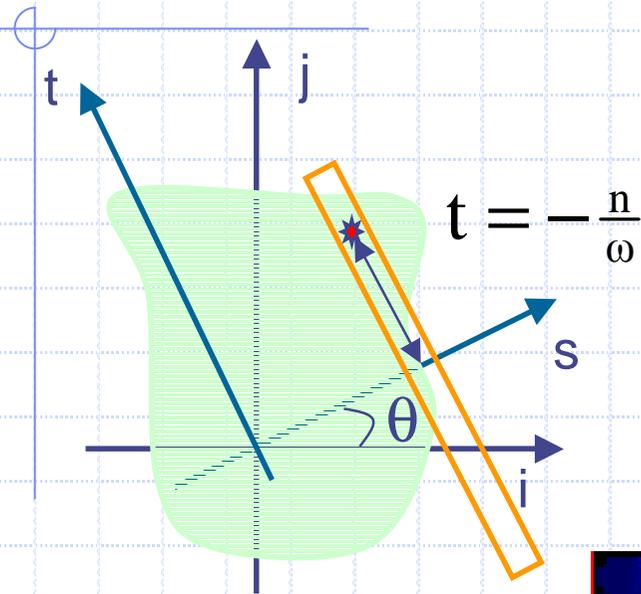
$$\mathbf{R} \cdot \vec{f} = \vec{p}$$



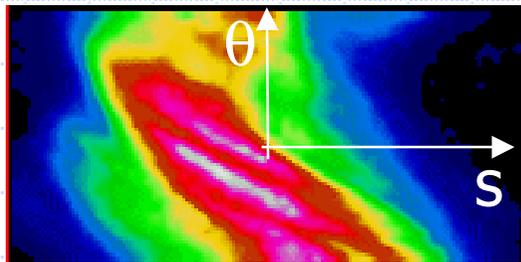
$${}^t\mathbf{R} \cdot \vec{p} = \vec{b}$$



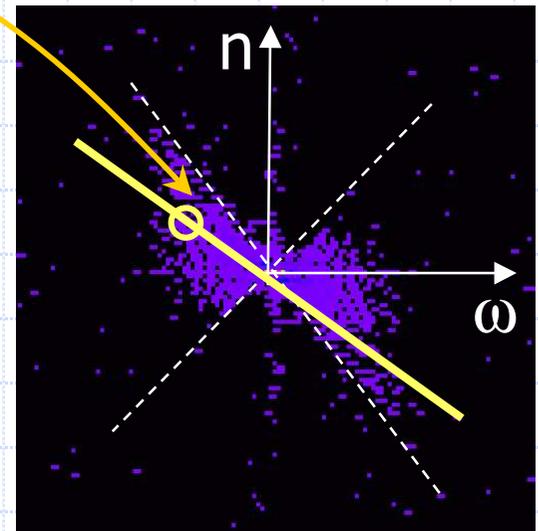
Déconvolution (hors synchro ECG)



sources à t mm du centre de rotation
 ↓
 signal \approx sur la droite $n = -\omega \cdot t$ de la TF2 du sinogramme



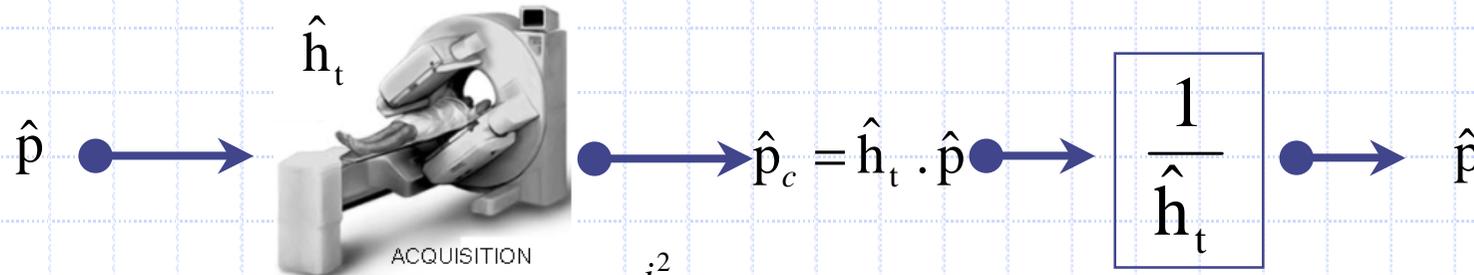
TF₂



$$p_c(s, \theta) = \int f(i, j) \cdot dt$$

$$\hat{p}_c(\omega, n)$$

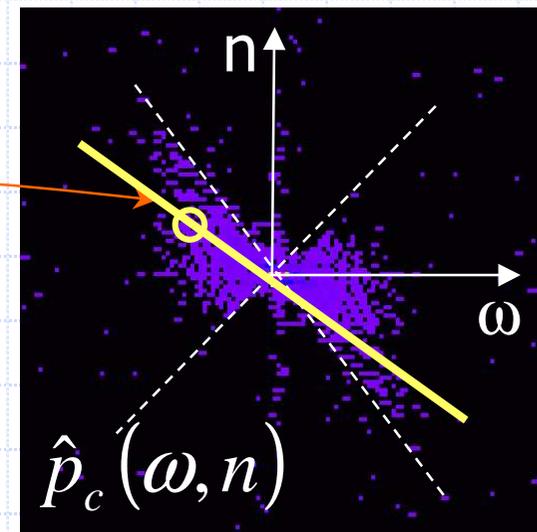
Principe fréquence-distance



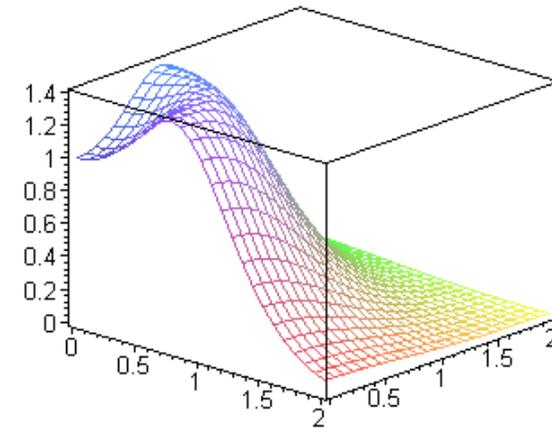
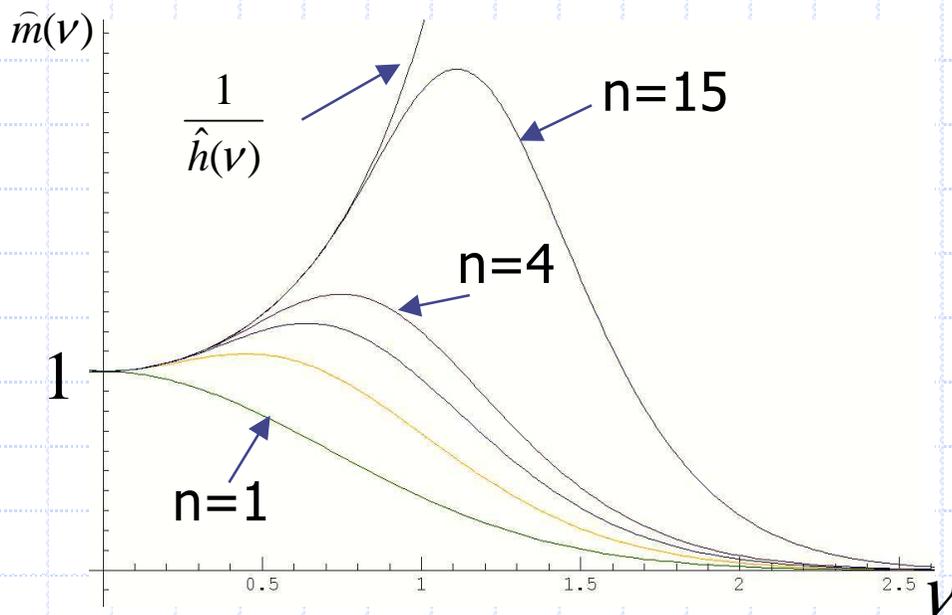
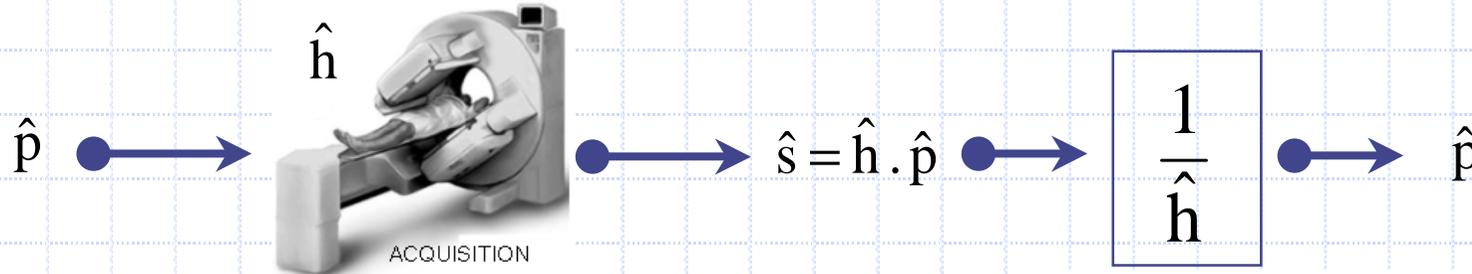
$$h_t(i) = \frac{1}{\sqrt{2\pi\sigma_t}} e^{-\frac{i^2}{2\sigma_t^2}}$$

$$\hat{p}_c(\omega, n) = \hat{h}_{\frac{n}{\omega}}(\omega) \cdot \hat{p}(\omega, n)$$

$$\hat{p}(\omega, n) = \frac{1}{\hat{h}_{\frac{n}{\omega}}(\omega)} \cdot \hat{p}_c(\omega, n)$$



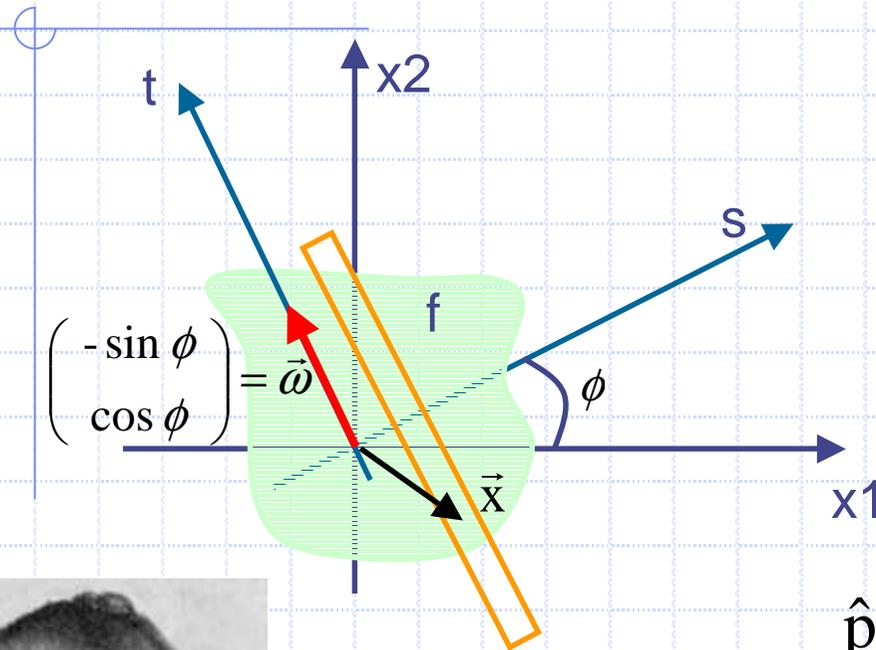
Filtre de déconvolution de Metz



$$\hat{m}(v, v') = \frac{1 - \left[1 - \hat{h}(v, v')^2 \right]^n}{\hat{h}(v, v')}$$

$$n = 0,834 \cdot \ln(C) - 7,774$$

Théorème de la projection



$$\begin{cases} p_{\vec{\omega}}(s) = \int_t f(s\vec{\omega}^\perp + t\vec{\omega}) dt \\ \hat{p}_{\vec{\omega}}(\sigma) = \int_s p_{\vec{\omega}}(s) \cdot e^{-i.s.\sigma} ds \end{cases}$$

$$\hat{p}_{\vec{\omega}}(\sigma) = \int_s \int_t f(s\vec{\omega}^\perp + t\vec{\omega}) e^{-i.s.\sigma} dt ds$$

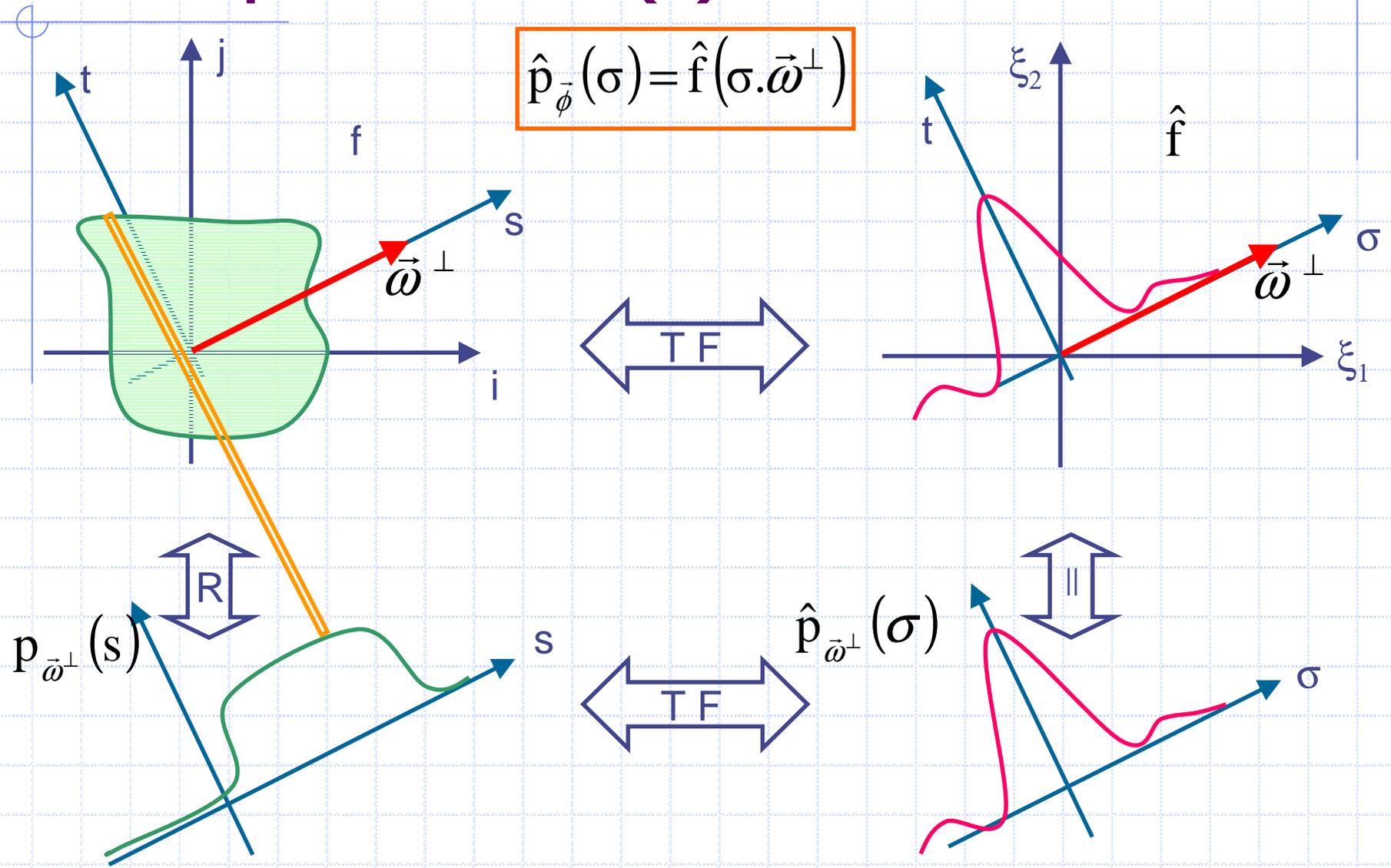
$$\hat{p}_{\vec{\omega}}(\sigma) = \iint f(\vec{x}) e^{-i.\sigma \vec{x} . \vec{\omega}^\perp} d\vec{x}$$

$$\hat{p}_{\vec{\omega}}(\sigma) = \hat{f}(\sigma.\cos\phi, \sigma.\sin\phi) = \hat{f}(\sigma.\vec{\omega}^\perp)$$

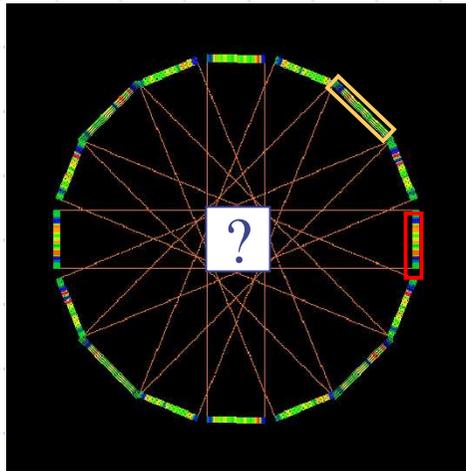


J. Radon
1887-1956

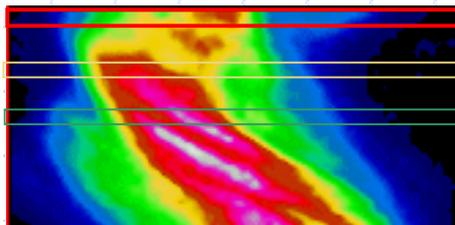
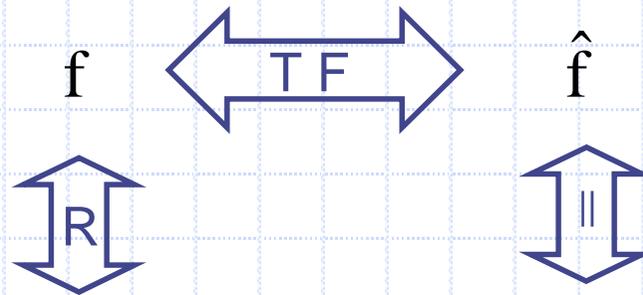
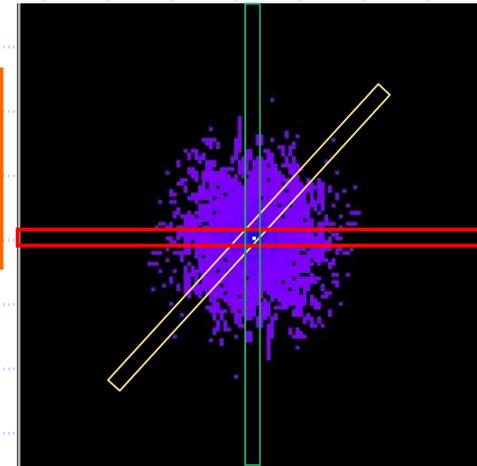
Interprétation (I)



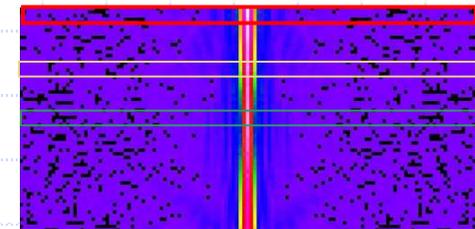
Interprétation (II)



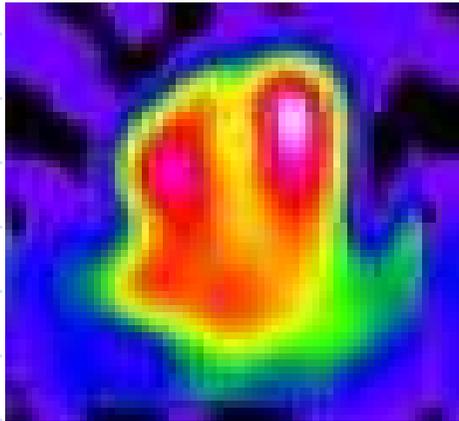
$$\hat{p}_{\vec{\omega}^\perp}(\sigma) = \hat{f}(\sigma \cdot \vec{\omega}^\perp)$$



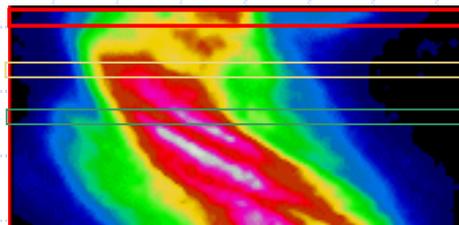
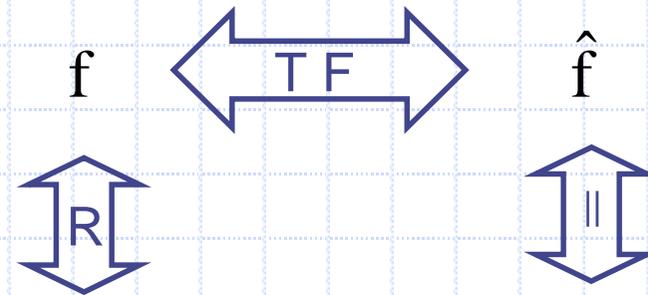
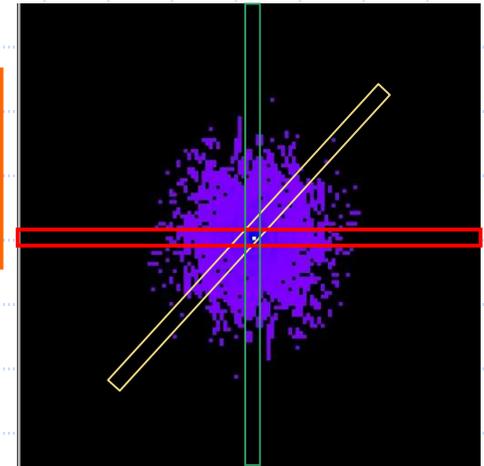
$$p_{\vec{\omega}^\perp}(s) \xleftrightarrow{\text{TF}} \hat{p}_{\vec{\omega}^\perp}(\sigma)$$



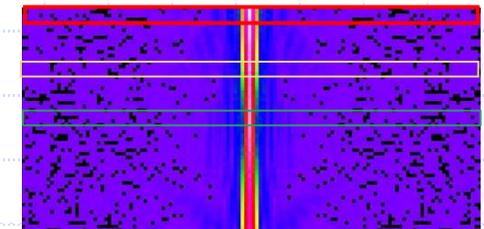
Interprétation (II)



$$\hat{p}_{\vec{\omega}^\perp}(\sigma) = \hat{f}(\sigma \cdot \vec{\omega}^\perp)$$



$$p_{\vec{\omega}^\perp}(s) \xleftrightarrow{\text{TF}} \hat{p}_{\vec{\omega}^\perp}(\sigma)$$



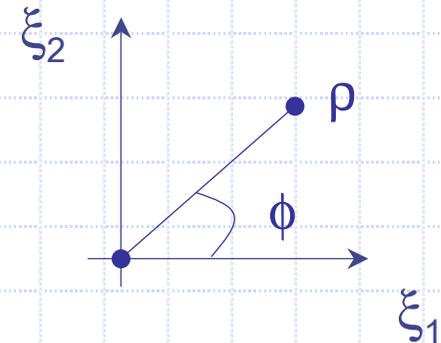
Rétroprojection filtrée (I)

$$f(\vec{x}) = \iint \hat{f}(\vec{\xi}) e^{i\vec{x} \cdot \vec{\xi}} d\vec{\xi}$$

$$f(\vec{x}) = \int_{\phi=0}^{\pi} \int_{\sigma=-\infty}^{\sigma=+\infty} \hat{f}(\sigma \vec{\omega}^{\perp}) e^{i\sigma \vec{\omega}^{\perp} \cdot \vec{x}} |\sigma| d\sigma d\phi$$

$$f(\vec{x}) = \int_{\phi=0}^{\pi} \int_{\sigma=-\infty}^{\sigma=+\infty} \hat{p}_{\vec{\omega}^{\perp}}(\sigma) |\sigma| e^{i\sigma \vec{\omega}^{\perp} \cdot \vec{x}} d\sigma d\phi$$

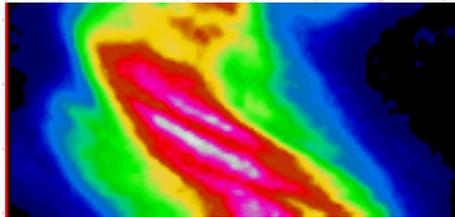
$$\underbrace{\text{TF}_s^{-1}[\hat{p}_{\vec{\omega}^{\perp}} \cdot \text{abs}]}_{p_{\vec{\omega}}^f}(\vec{\omega}^{\perp} \cdot \vec{x})$$



J. Radon
1887-1956

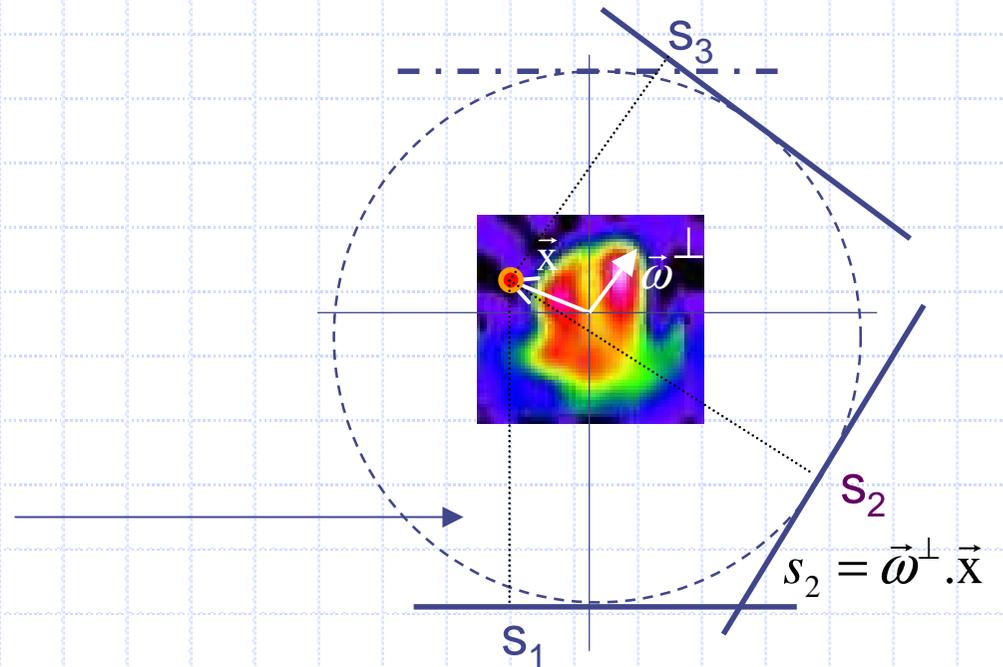
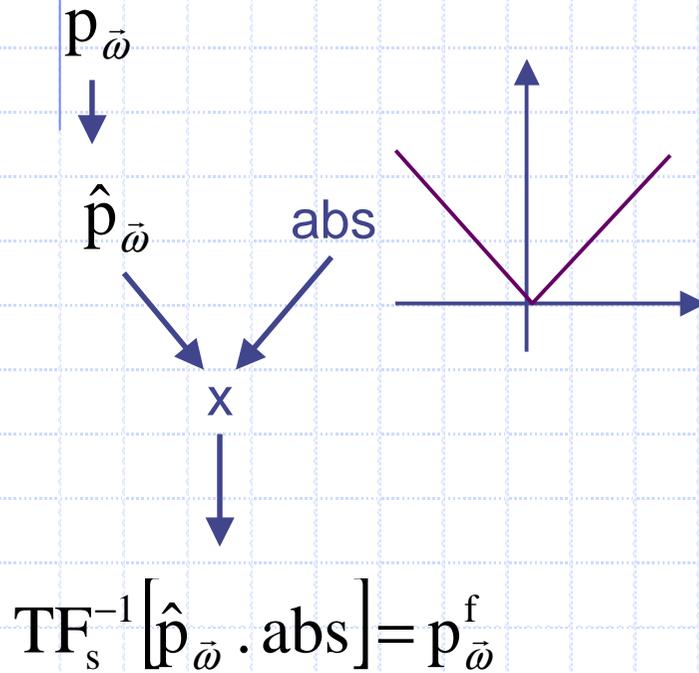
$$f(\vec{x}) = (\mathbf{R}^* p^f)(\vec{x})$$

Rétroprojection filtrée (II)

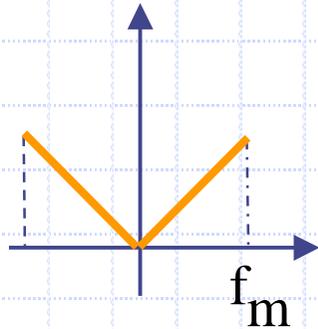


$$f(\vec{x}) = (R^* p^f)(\vec{x})$$

Projections sur 180°



Rétroprojection filtrée (III)



$$f_m = \frac{1}{2 \cdot \Delta r} = \frac{f_e}{2}$$

$$RL(x) = \frac{f_m \sin(2\pi f_m x)}{\pi x} - \frac{1 - \cos(2\pi f_m x)}{2\pi^2 x^2}$$

$$RL(k \cdot \Delta r) = \begin{cases} \frac{1}{4(\Delta r)^2} & k = 0 \\ 0 & k \neq 0 \text{ pair} \\ -\frac{1}{(k\pi \Delta r)^2} & k \text{ impair} \end{cases}$$

Exemple pour $\Delta r = \frac{1}{2}$, on obtient le filtre : $\left(0 \quad -\frac{1}{2,5} \quad 1 \quad -\frac{1}{2,5} \quad 0 \right)$

Rétroprojection filtrée (IV)

0	0	0	← 15 ← 45
0	0	0	← 60 ← 90
0	0	0	← 15 ← 45

45 90 45

$$\text{Filtre} = \begin{pmatrix} -\frac{1}{3} & 1 & -\frac{1}{3} \end{pmatrix}$$

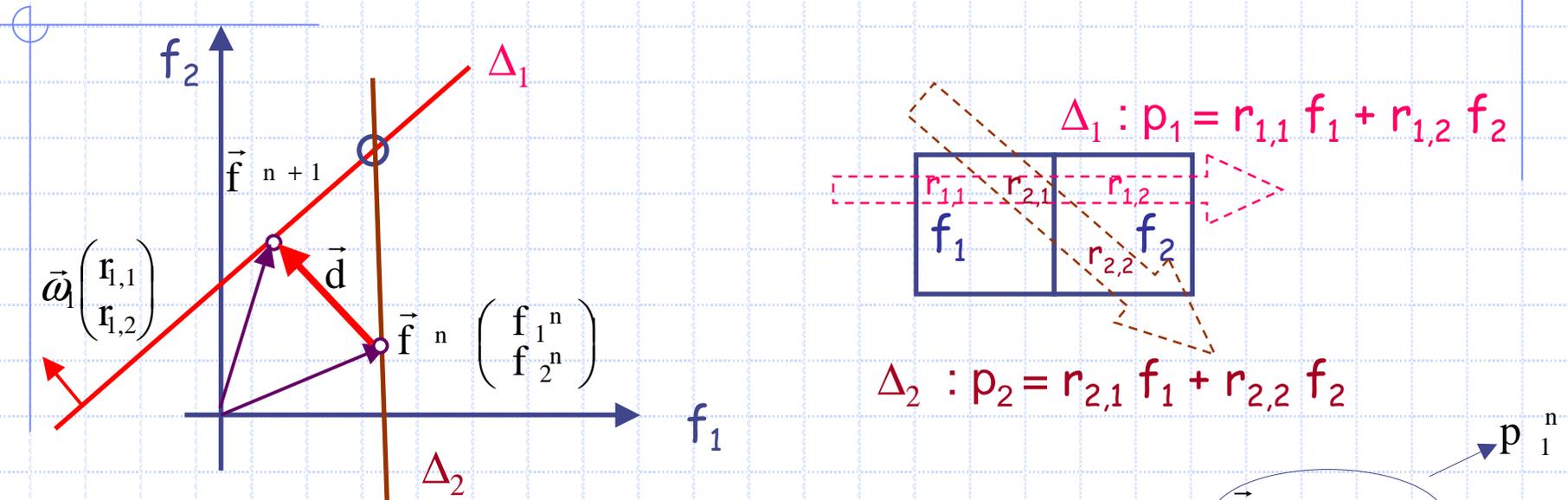
10	25	10
25	40	25
10	25	10

5	5	5
20	20	20
5	5	5
↑ 15	↑ 60	↑ 15

⇒



Algebraic Reconstruction Technique (I)



$$\vec{f}^{n+1} = \vec{f}^n + d \frac{\vec{\omega}_1}{\|\vec{\omega}_1\|} \quad d = \frac{p_1 - \vec{f}^n \cdot \vec{\omega}_1}{\|\vec{\omega}_1\|}$$

$$\vec{f}^{n+1} = \vec{f}^n + \frac{p_1 - p_1^n}{\|\vec{\omega}_1\|^2} \vec{\omega}_1$$

$$\vec{f}^{n+1} = \vec{f}^n + R^* (p_1 - p_1^n)$$



S. Kaczmarz
1895-1940

Algebraic Reconstruction Technique (II)

0	0	0
0	0	0
0	0	0

$$\leftarrow 45 - 0 = 15 + 15 + 15$$

$$\leftarrow 90 - 0 = 30 + 30 + 30$$

$$\leftarrow 45 - 0$$

45 90 45

⇒

15	15	15
30	30	30
15	15	15

↓ ↓ ↓
45 90 45

- 60 60 60

-15 30 -15

10	25	10
25	40	25
10	25	10

$$\vec{f}^{n+1} = \vec{f}^n + R * (p_1 - p_1^n)$$

MLEM et OSEM

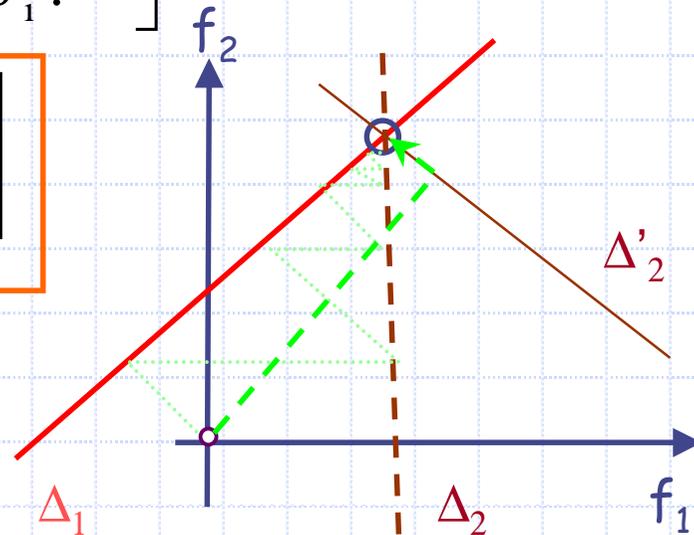
Idée: $\text{argmax}_{\vec{f}} P(\vec{f} / \vec{p}') & \text{stat. de Poisson sur } p$

Bayes: $P(\vec{f} / \vec{p}') = P(\vec{p}' / \vec{f}) \cdot P(\vec{f}) / P(\vec{p}') = P(\vec{p}' / \vec{f}) \cdot P(\vec{f})$

$$\vec{f} \approx \text{arg min}_{\vec{f}} \left[-\log P(\vec{p}' / \vec{f}) \right] = \left[-\log \prod_{i=1}^K \frac{e^{-p_i} p_i^{p'_i}}{p'_i!} \right]$$

$$\vec{f}^{n+1} = \vec{f}^n \cdot \frac{1}{\sum_{l'} r_{l',i}} R^* \begin{bmatrix} p_i \\ p_i^n \end{bmatrix}$$

$$f_i^{n+1} = f_i^n \cdot \frac{1}{\sum_{l'=1}^P r_{l',i}} \sum_{l=1}^P r_{l,i} \frac{p_l}{\sum_{s=1}^N r_{l,s} f_s^n}$$



Gradient Conjugué

$$\bar{f} = \arg \min_{f \in C} \| R\vec{f} - \vec{p} \|^2$$

Initialisation :

$$\vec{d}^0 = \vec{r}^0 = R^* \cdot \vec{p} \longrightarrow$$

$$\omega^j = \frac{\|\vec{r}^j\|^2}{\langle \vec{d}^j | R^* \cdot R \cdot \vec{d}^j \rangle}$$

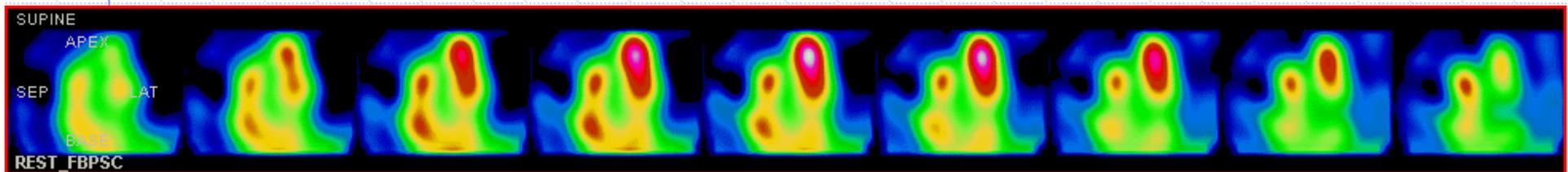
$$\vec{r}^{j+1} = \vec{r}^j - \omega^j \cdot R^* \cdot R \cdot \vec{d}^j$$

$$\vec{d}^{j+1} = \vec{r}^{j+1} + \frac{\|\vec{r}^{j+1}\|^2}{\|\vec{r}^j\|^2} \cdot \vec{d}^j$$

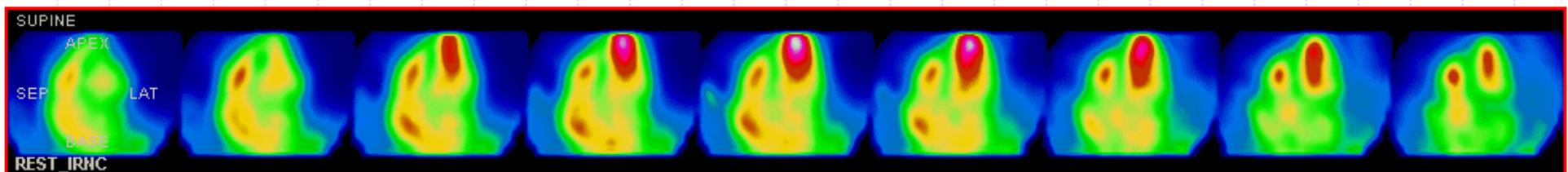
$$\vec{f}^{j+1} = \vec{f}^j + \omega^j \cdot \vec{d}^j$$

Exemples

◆ RPF + BTW 10, 20, 40, 80, 100 % Nyquist



◆ MLEM 10, 50 et 100 itérations



Problème d'Hadamard bien posé ?

✓ En continu : $\hat{p}_{\vec{\phi}}(\sigma) = \hat{f}(\sigma \cdot \vec{\omega}^\perp)$, R bijectif d'inverse continue (conditions d'Hadamard).

✓ En discret, les choses sont moins simples :

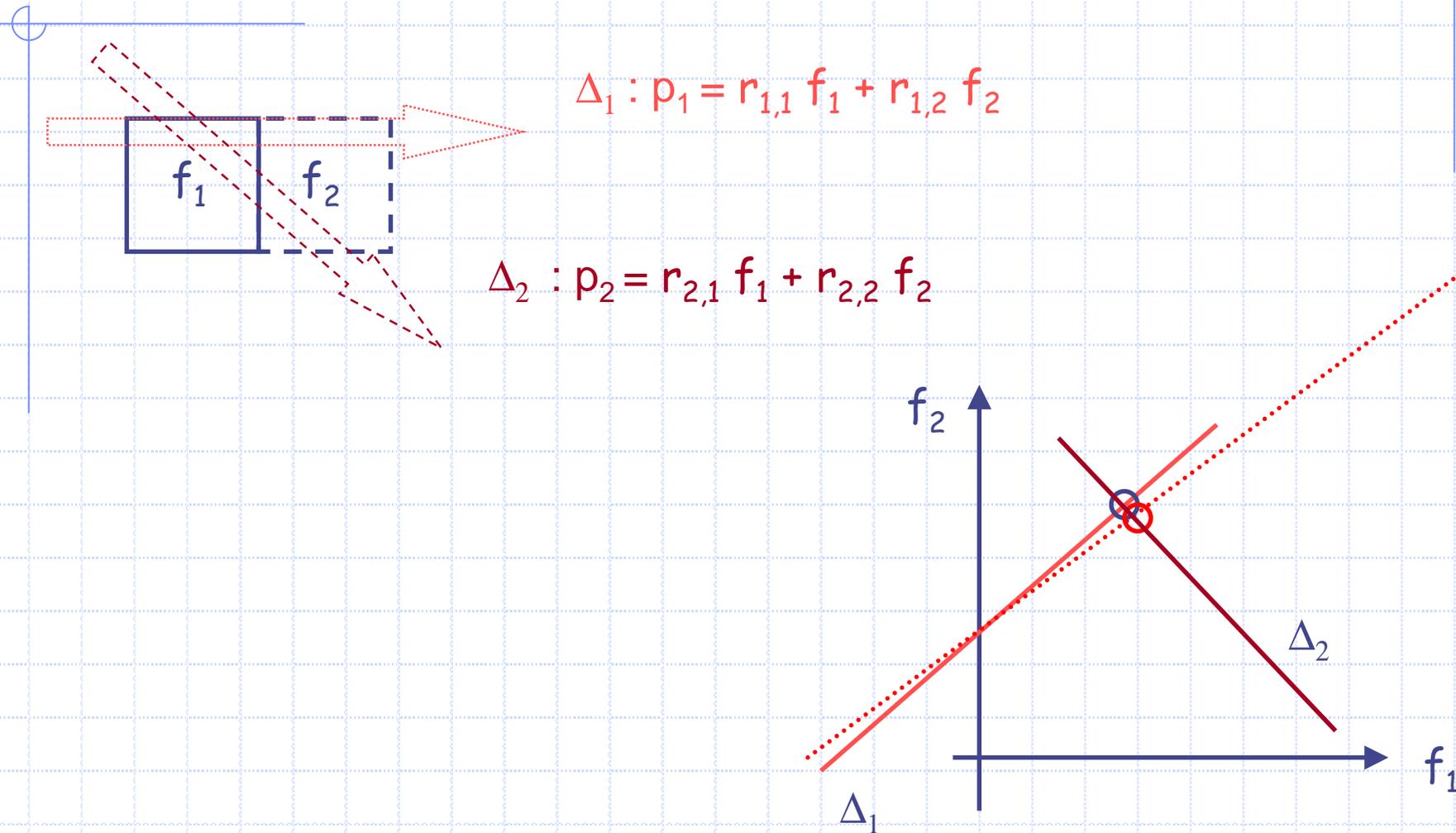
- R surjectif ? $\iff {}^t R \cdot R \vec{f} = A \vec{f} = {}^t R \cdot \vec{p} = \vec{q} \iff \vec{f} = \arg \min_{f \in C} \|\vec{p} - R \vec{f}\|^2$

- R injectif ? : choix parmi les solutions possibles

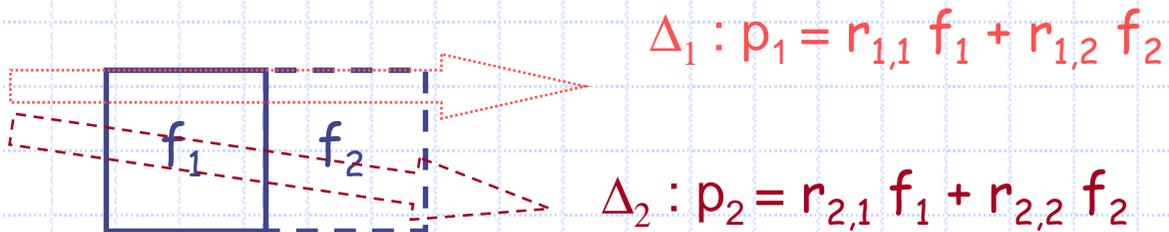
- R^{-1} continue mais $\|R^{-1}\|$ grande : $\kappa(R) = \|R\| \|R^{-1}\| = \frac{\lambda_{\max}}{\lambda_{\min}} \gg 1$

$$\frac{\|\delta \vec{f}\|}{\|\vec{f}\|} \leq \frac{\kappa(R)}{1 - \kappa(R) \frac{\|\delta R\|}{\|R\|}} \left[\frac{\|\delta \vec{p}\|}{\|\vec{p}\|} + \frac{\|\delta R\|}{\|R\|} \right]$$

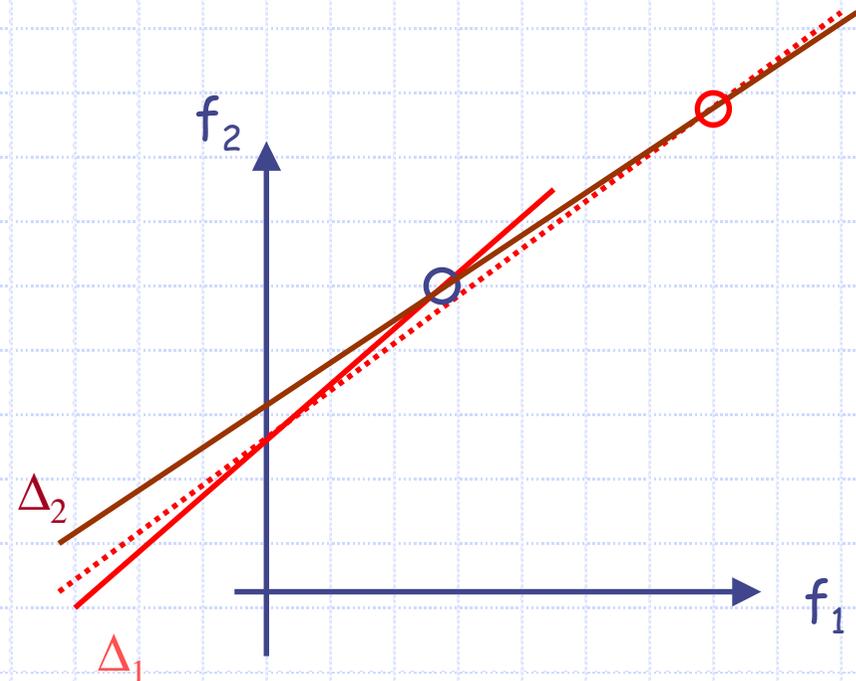
Approche intuitive (I)



Approche intuitive (II)



$64^2 = 4\ 096$
 $128^2 = 16\ 384$
 $256^2 = 65\ 536$
 $512^2 = 262\ 144$



Exemple

Matrice de Wilson

$$\begin{pmatrix} 10 & 7 & 8 & 7 \\ 7 & 5 & 6 & 5 \\ 8 & 6 & 10 & 9 \\ 7 & 5 & 9 & 10 \end{pmatrix} \begin{pmatrix} 1 \\ 1 \\ 1 \\ 1 \end{pmatrix} = \begin{pmatrix} 32 \\ 23 \\ 33 \\ 31 \end{pmatrix}$$

$$\text{Sp}(R) \approx \{0,010; 0,843; 3,858; 30,289\} \Rightarrow \kappa(R) = \frac{30,289}{0,010} = 2984$$

$$\begin{pmatrix} 10 & 7 & 8 & 7 \\ 7 & 5 & 6 & 5 \\ 8 & 6 & 10 & 9 \\ 7 & 5 & 9 & 10 \end{pmatrix} \begin{pmatrix} 9,2 \\ -12,6 \\ 4,5 \\ -1,1 \end{pmatrix} = \begin{pmatrix} 32,1 \\ 22,9 \\ 33,1 \\ 30,9 \end{pmatrix}$$

$$\begin{pmatrix} 10 & 7 & 8 & 7 \\ 7 & 5 & 6 & 5 \\ 8 & 6 & 10 & 9 \\ 7 & 5 & 9 & 10 \end{pmatrix} \begin{pmatrix} -7,2 \\ 14,6 \\ -2,5 \\ 3,1 \end{pmatrix} = \begin{pmatrix} 31,9 \\ 23,1 \\ 32,9 \\ 31,1 \end{pmatrix}$$

Estimation de $\kappa(R)$

◆ Gradient conjugué

◆ Approximation de Galerkin

$$\omega^j = \frac{\|\vec{r}^j\|^2}{\langle \vec{d}^j | R^* \cdot R \cdot \vec{d}^j \rangle} \quad , \quad \beta^j = \frac{\|\vec{r}^{j+1}\|^2}{\|\vec{r}^j\|^2}$$

$$\begin{pmatrix} \frac{1}{\omega^0} & -\frac{\sqrt{\beta^0}}{\omega^0} & 0 & 0 \\ -\frac{\sqrt{\beta^0}}{\omega^0} & \frac{1}{\omega^1} + \frac{\beta^0}{\omega^0} & \ddots & 0 \\ 0 & \ddots & \ddots & -\frac{\sqrt{\beta^{j-1}}}{\omega^{j-1}} \\ 0 & 0 & -\frac{\sqrt{\beta^{j-1}}}{\omega^{j-1}} & \frac{1}{\omega^j} + \frac{\beta^{j-1}}{\omega^{j-1}} \end{pmatrix}$$

- ◆ dont le spectre CV vers celui de R: estimation de $\kappa(R)$
- ◆ qui borne l'erreur sur la solution et le spectre

Exemple de régularisation (I)

Remplacer : $\bar{f} = \arg \min_{f \in C} \|\vec{p} - R\vec{f}\|^2$ par

$$\bar{f} = \arg \min_{f \in C} \left\{ \|\vec{p} - R\vec{f}\|^2 + \alpha \cdot \rho(\vec{f}) \right\}$$

Adéquation aux données
Surjectivité du problème inverse

Régularisation
injectivité

Exemple : régularisation de Tikhonov (cf. pseudo-inverse de Moore-Penrose)

$$\vec{f} = \arg \min_{\vec{f}} \left\{ \|\vec{p} - R\vec{f}\|^2 + \alpha \cdot \|\vec{f}\|^2 \right\} \Leftrightarrow (R^t R + \alpha I) f = R^t p$$

$$\vec{f} = (R^t R + \alpha I)^{-1} R^t p$$

Exemple de régularisation (II)

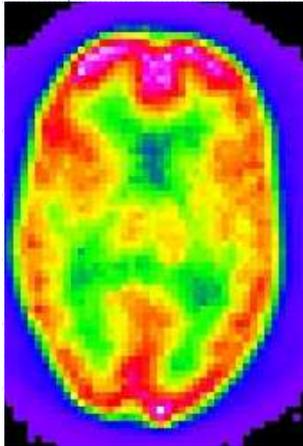
Bayes : $P(\vec{f} / \vec{p}) = P(\vec{p}/\vec{f}) \cdot P(\vec{f}) / P(\vec{p}) = P(\vec{p}/\vec{f}) \cdot P(\vec{f})$

$$\vec{f} \approx \arg \min_{\vec{f}} \left[\underbrace{-\log P(\vec{p}/\vec{f})}_{\text{Adéquation aux données}} - \underbrace{\log P(\vec{f})}_{\text{régularisation}} \right]$$

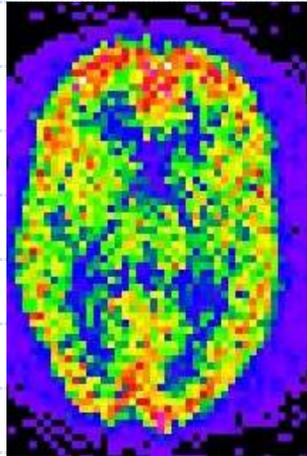
Distribution de Gibbs : $P(\vec{f}) = \frac{1}{K} e^{-\beta \cdot \sum_{i,j} w_{i,j} \cdot V(f_i - f_j)}$

$$\vec{f} \approx \arg \min_{\vec{f}} \left[-\log P(\vec{p}/\vec{f}) + \beta \sum_{i,j} w_{i,j} V(f_i - f_j) \right] \quad \text{MAP-EM-OSL}$$

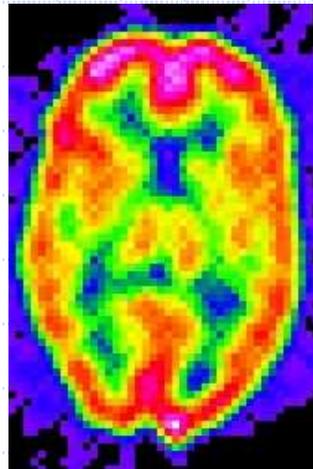
Comparaison MLEM-GC-FRECT



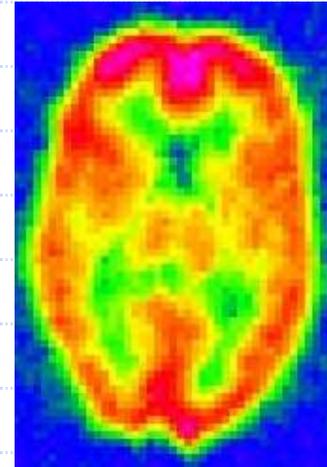
MLEM 6



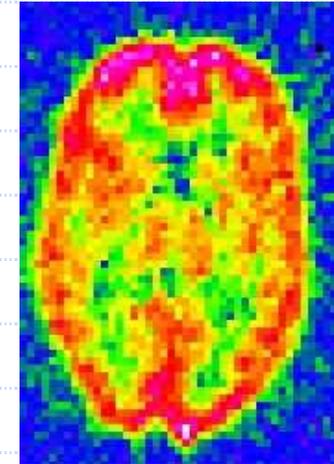
MLEM 200



FRECT 34 (CV)



GC 6



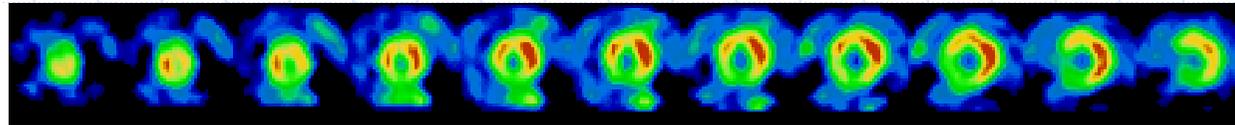
GC 16

$$E(f) = \left\| \text{TF}_s^{-1}(B \cdot \hat{p}) - Rf \right\|^2 + \left\| (1 - B) \cdot \hat{f} \right\|^2$$

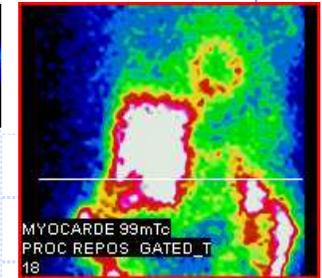
1. Problème médical à résoudre
2. Choix d'une modalité d'imagerie
3. Choix d'un traceur
4. Acquisition des images
5. Reconstruction tomographique
6. Segmentation, analyse et validation (myocarde)
7. Segmentation, analyse et validation (ventriculo)

Segmentation

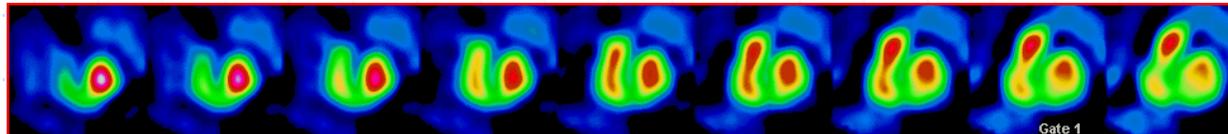
◆ Scintigraphie myocardique (MIBI ou MIBG)



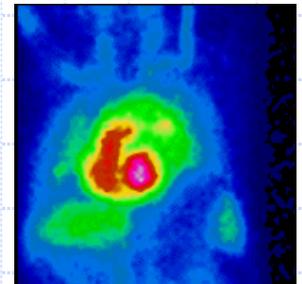
- ◆ Faible rapport signal/bruit
- ◆ Activités extra cardiaques
- ◆ Ventricule gauche \approx ellipsoïde tronqué



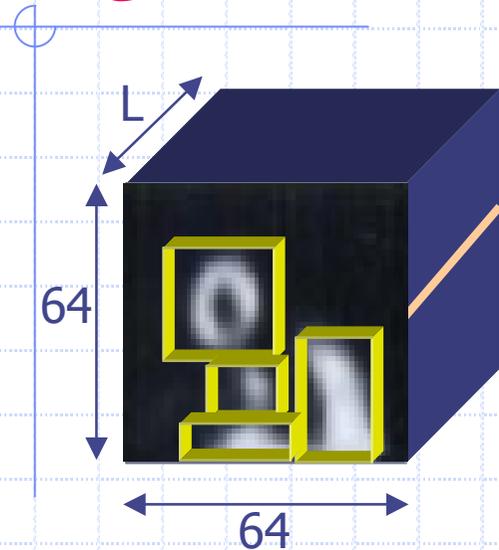
◆ Tomoventriculographie isotopique



- ◆ Meilleur rapport S/B, basse fréquence
- ◆ Volumes partiels
- ◆ Peu d'activité extra cardiaque
- ◆ Pas de modélisation simple (ventricule droit)



QPS[®] : MYOCARDE



S+5% ⇒ fragmentation

Essai d'identification :
seuil → 1 amas cubique [50 mL, 1L[

① Somme pondérée des coupes SA

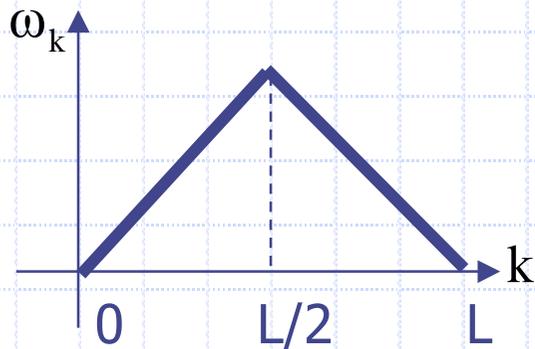
$$S(i, j) = \sum_{k=1}^L \omega_k \cdot SA(i, j, k)$$

② ? Maxima locaux de S(i,j)

- supérieur seuil et au moins 4-connexes

③ Transformée de Hough

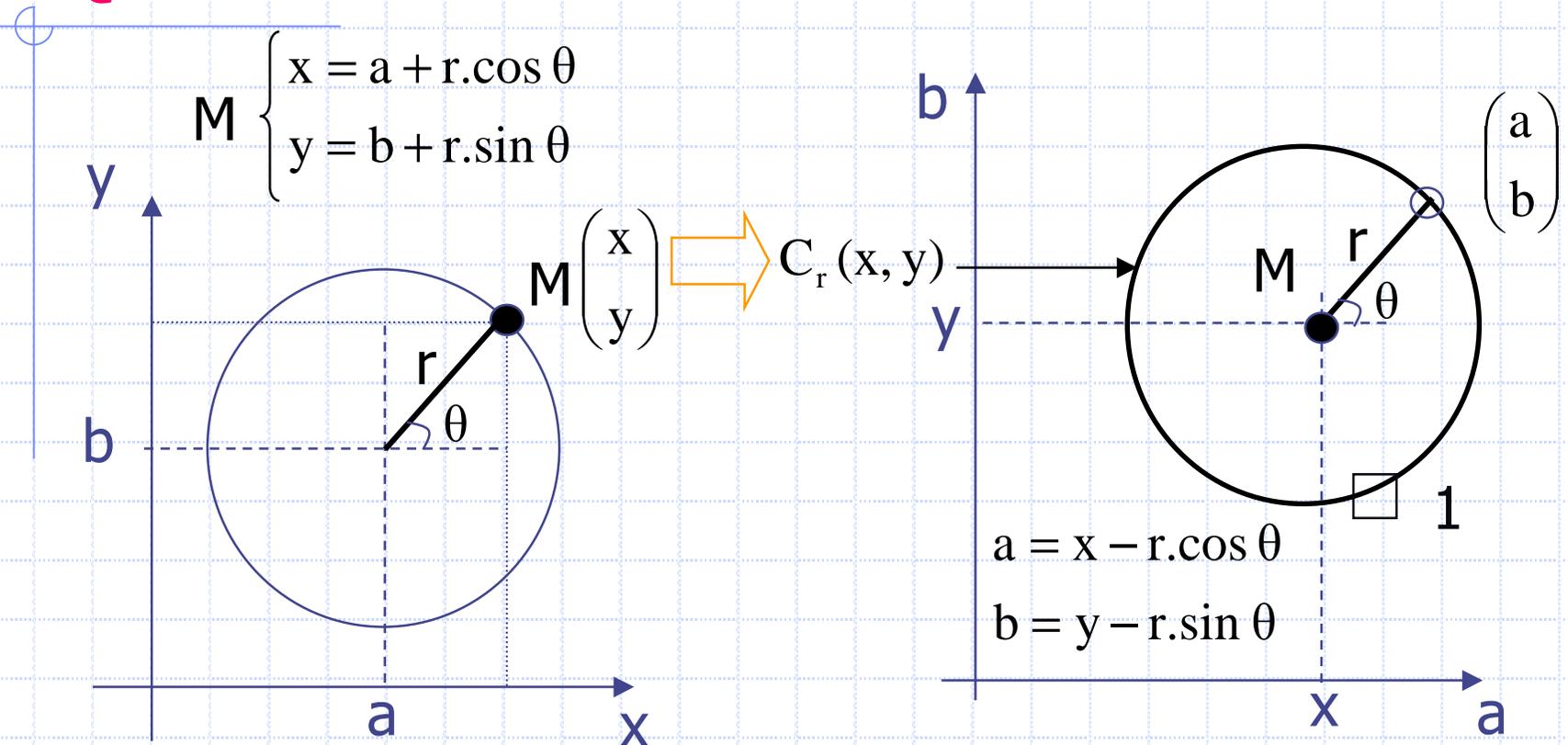
- ? cercles dans un nuage de points



A Ezekiel et al. in Computers in cardiology. NY, IEEE Computer society. 1991:237-240

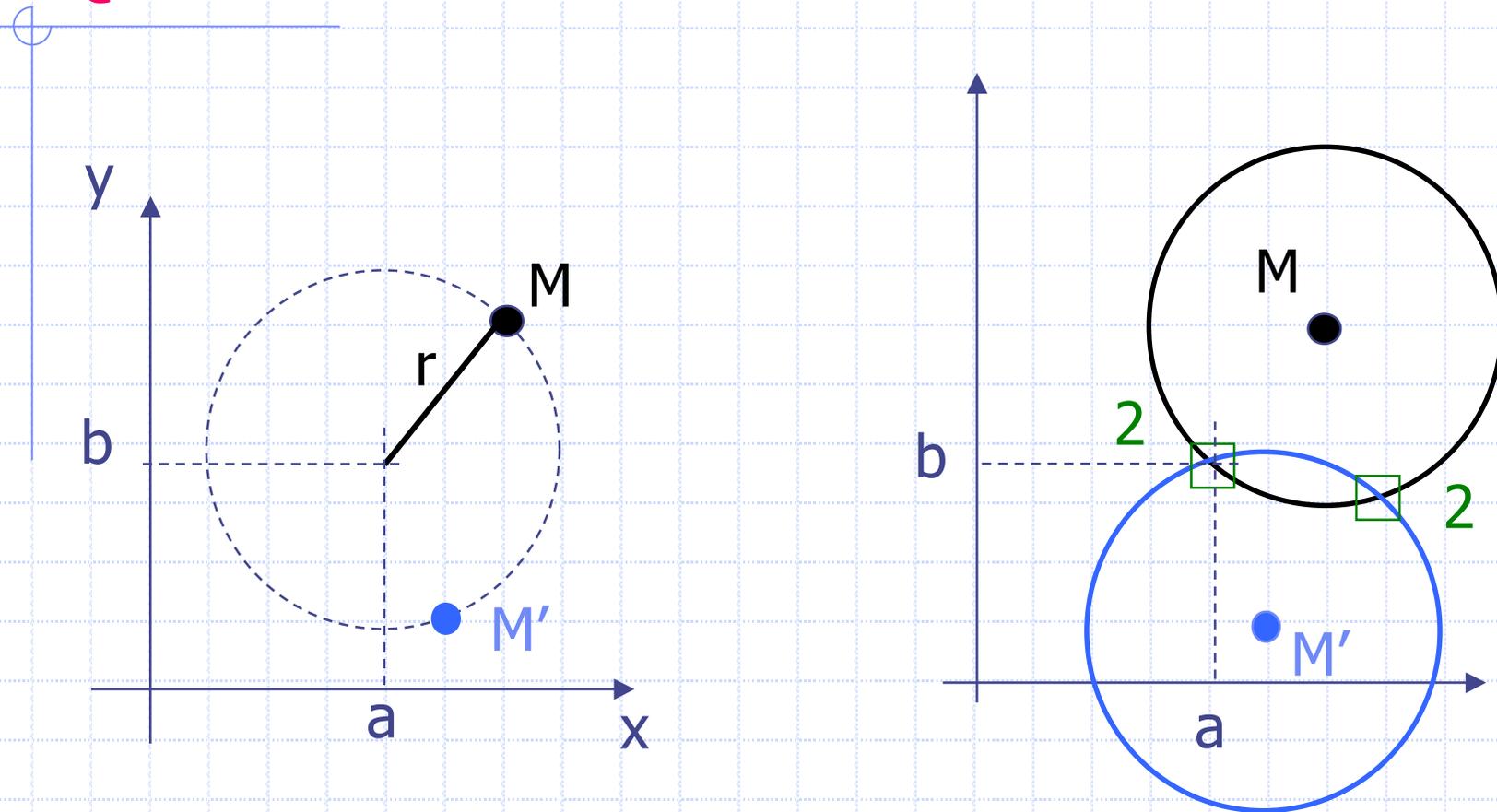
G. Germano et al. JNM 1995;36:1107-1114 et 1995;36:2138-2147

QPS[®] : TRANSFORMATION DE HOUGH



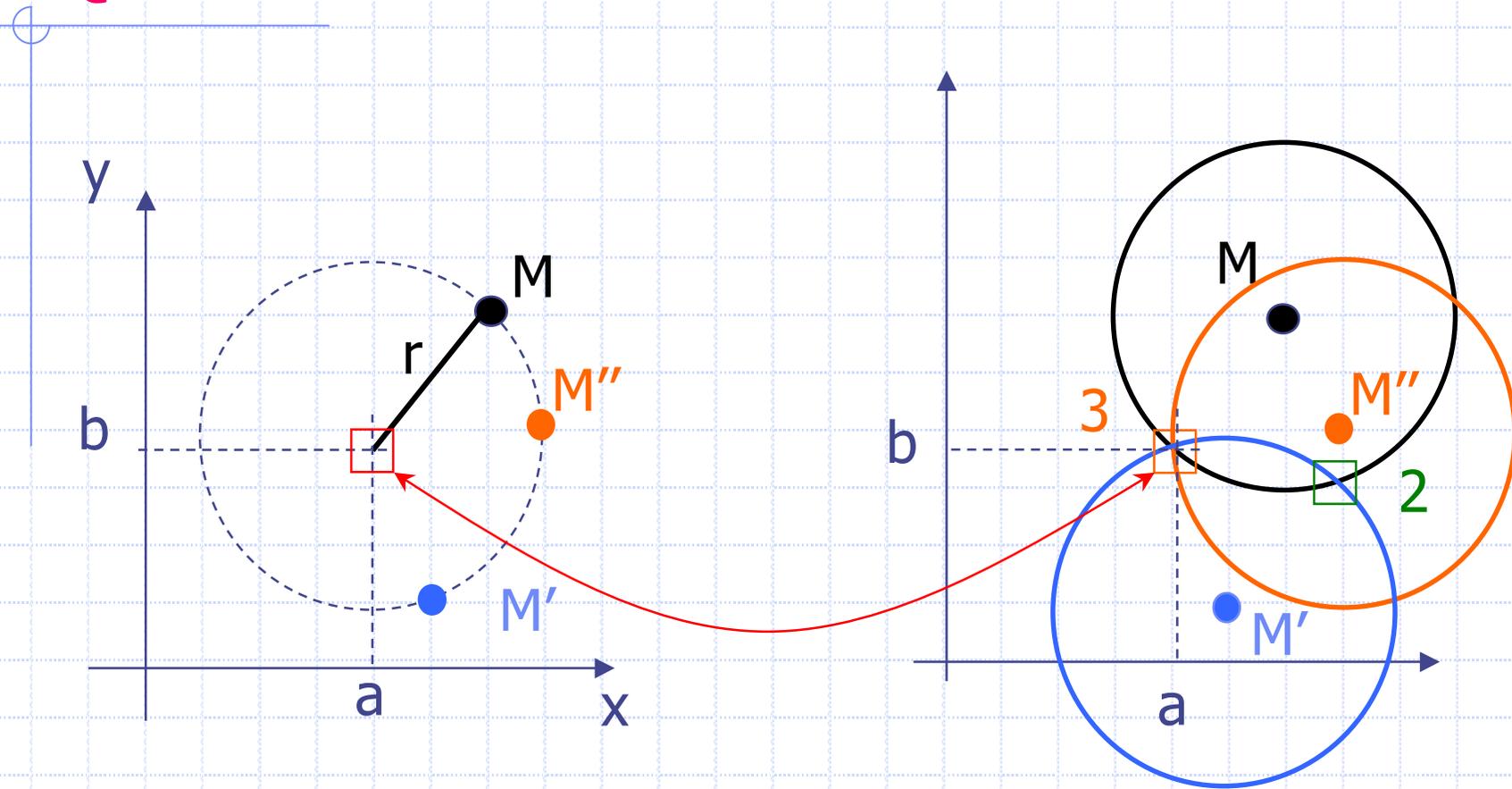
à tout **point** M du cercle (O, r) on associe un **cercle** (M, r)

QPS[®] : TRANSFORMATION DE HOUGH



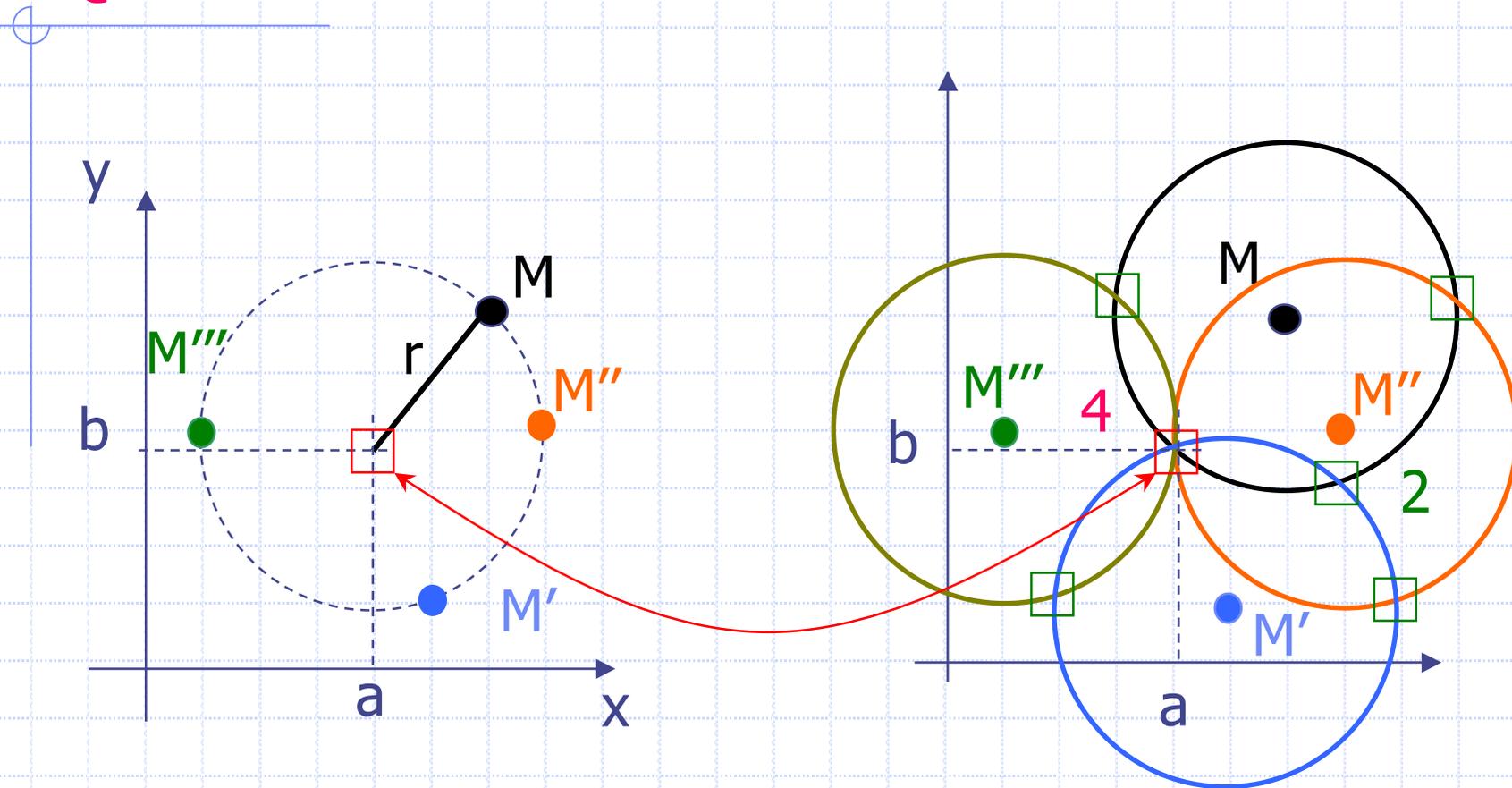
D. Ballard, C. Brown. Computer vision. Englewood Cliffs, NJ: Prentice-Hall; 1982.
<http://homepages.inf.ed.ac.uk/rbf/BOOKS/BANDB/bandb.htm>

QPS[®] : TRANSFORMATION DE HOUGH



D. Ballard, C. Brown. Computer vision. Englewood Cliffs, NJ: Prentice-Hall; 1982.
<http://homepages.inf.ed.ac.uk/rbf/BOOKS/BANDB/bandb.htm>

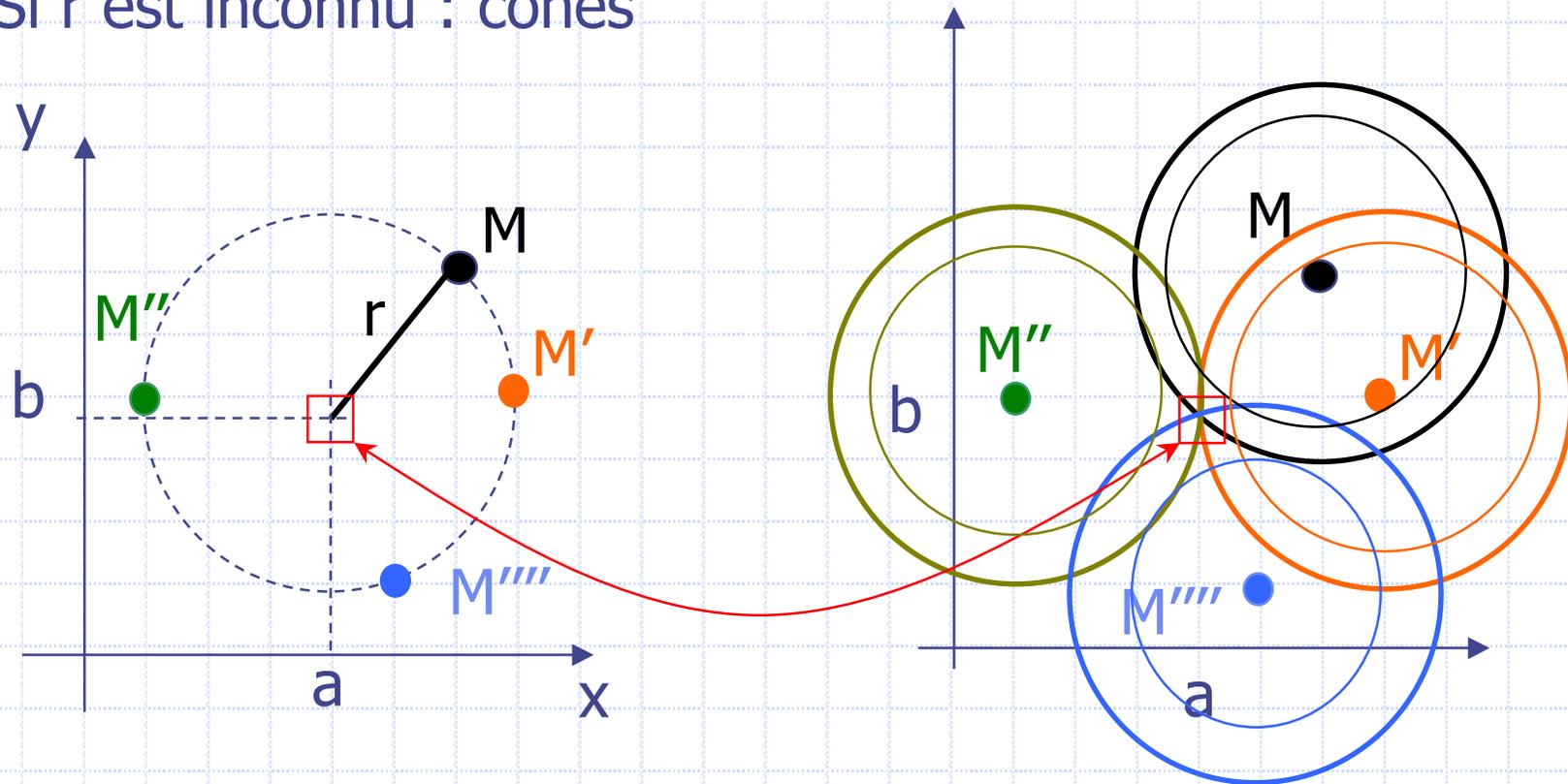
QPS[®] : TRANSFORMATION DE HOUGH



D. Ballard, C. Brown. Computer vision. Englewood Cliffs, NJ: Prentice-Hall; 1982.
<http://homepages.inf.ed.ac.uk/rbf/BOOKS/BANDB/bandb.htm>

QPS[®] : TRANSFORMATION DE HOUGH

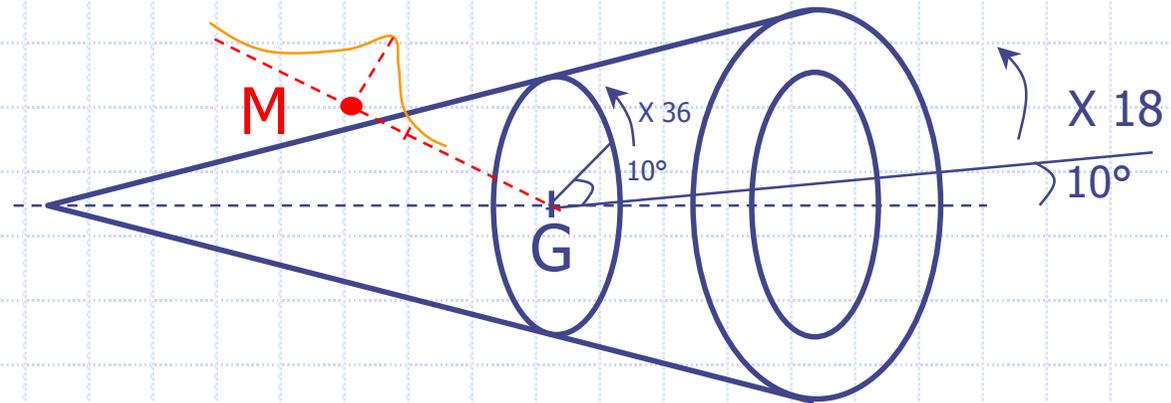
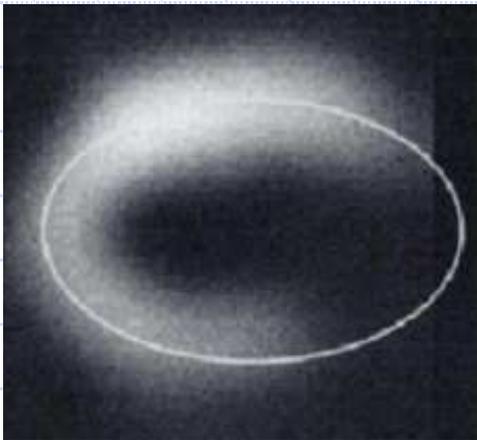
Si r est inconnu : cônes



? Extrema dans l'espace des paramètres 3D: (a, b, r)
Puis QPS choisit le cercle à plus fort comptage moyen sur la circonférence par rapport au comptage au centre

QPS[®] : MYOCARDE

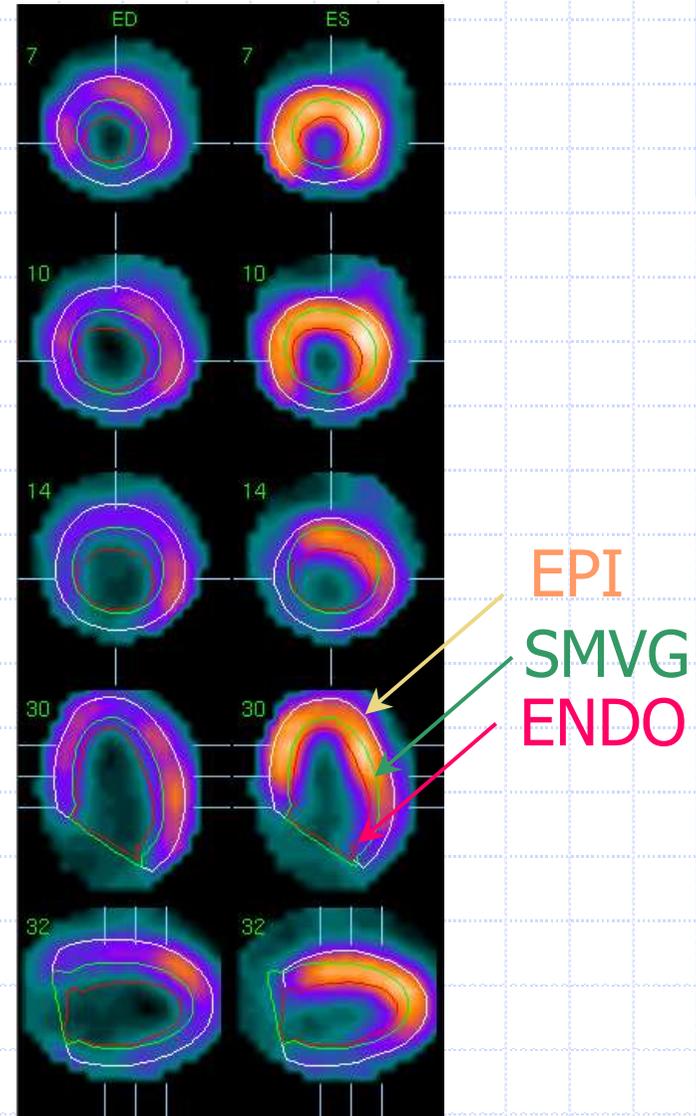
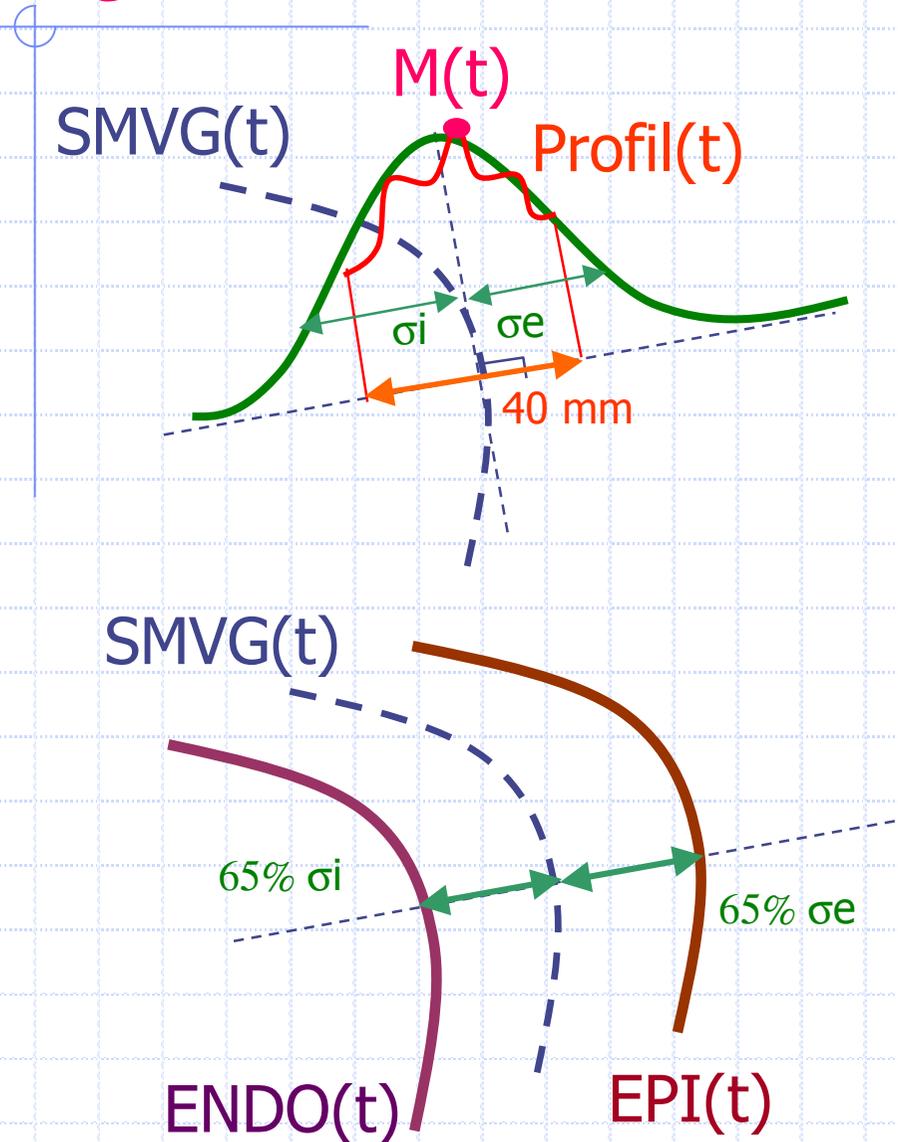
- ① G : Centre de Gravité du masque binaire
- ② Données = Volume SPECT x Masque
- ③ Extraction de 18x36 profils d'activité
- ④ Surface médio VG = maxima des profils M
- ⑤ Fit ellipse 3D → axe VG → G ← proj.(G)/axe



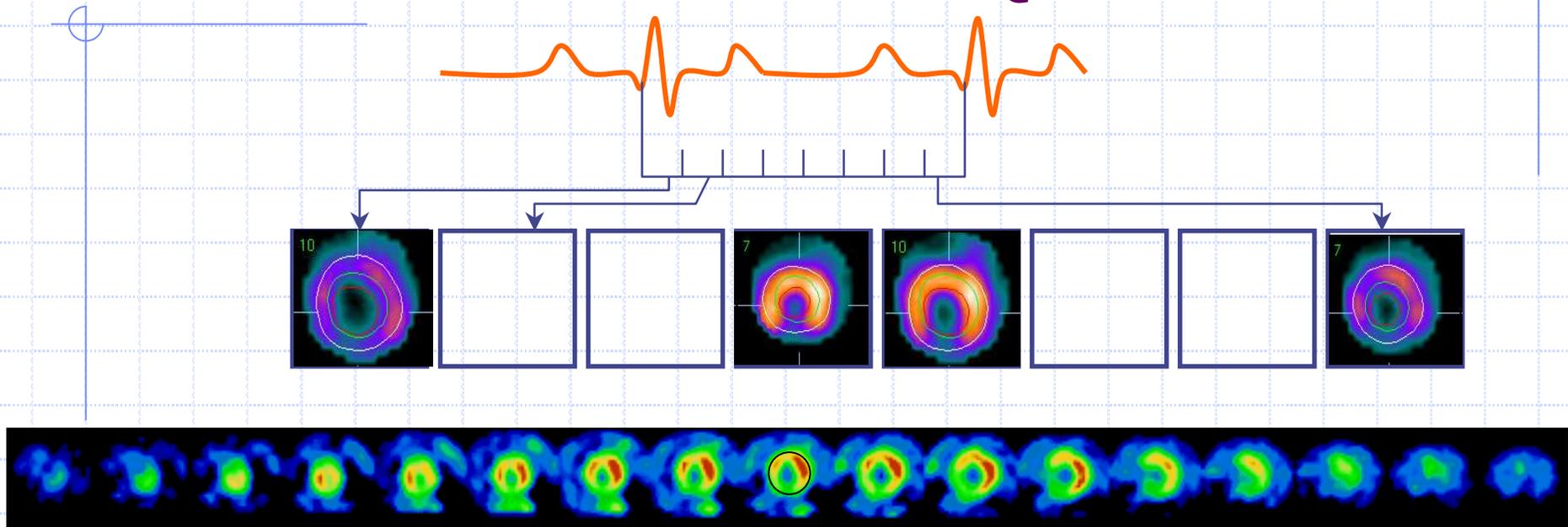
JC Cauvin et al. Eur J Nucl med 1992;19:1032-1037

G. Germano et al. JNM 1995;36:1107-14 ; 1995;36:2138-47 et JNM 2000;41:712-19

QPS[®] : SEGMENTATION DU MYOCARDE

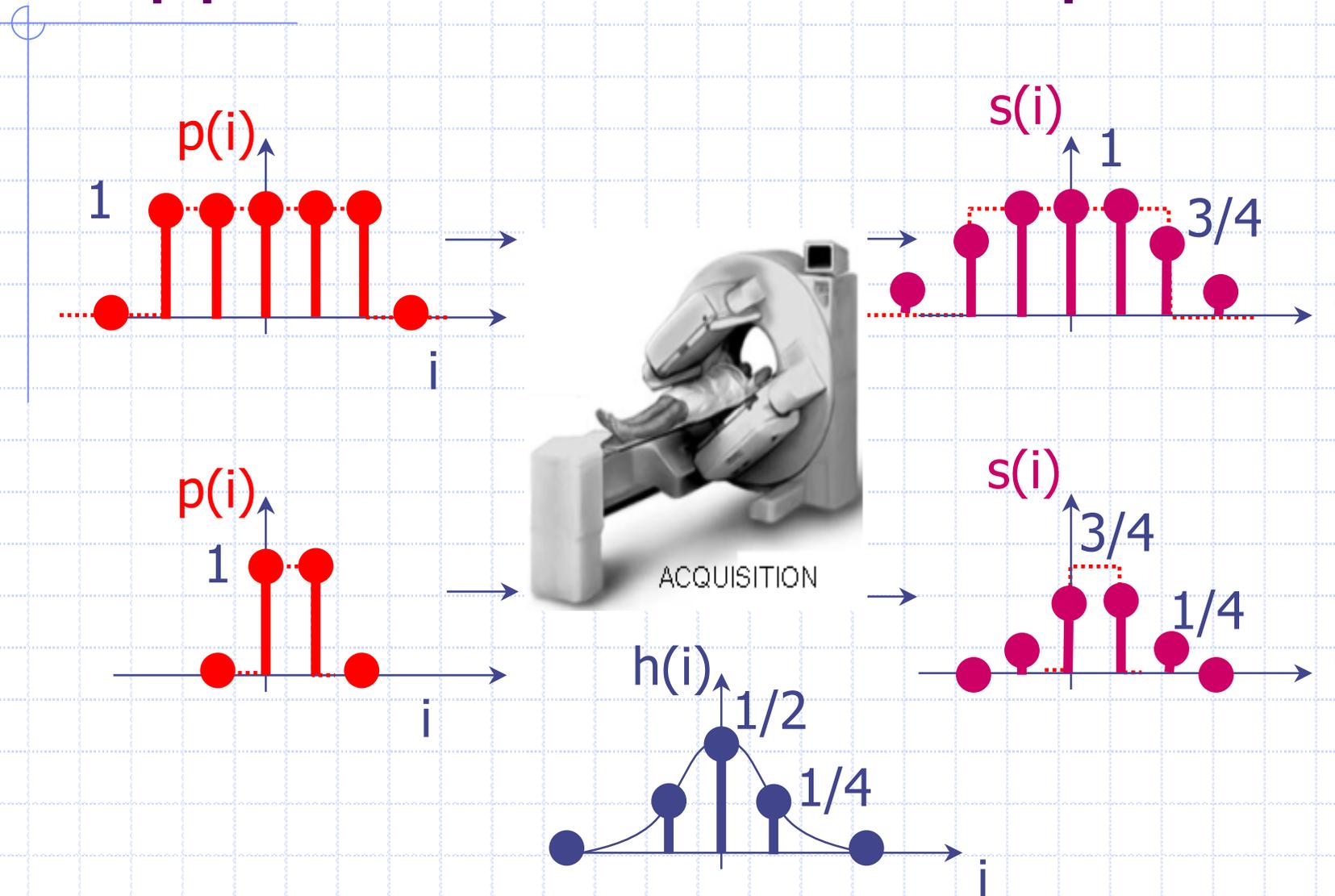


MOUVEMENT MYOCARDIQUE ?

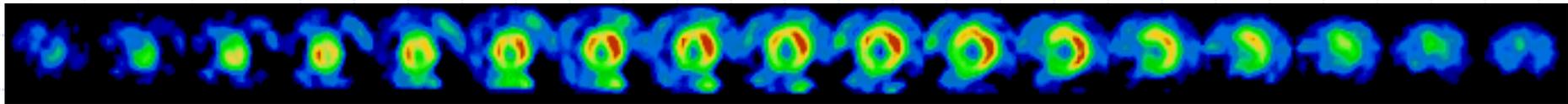
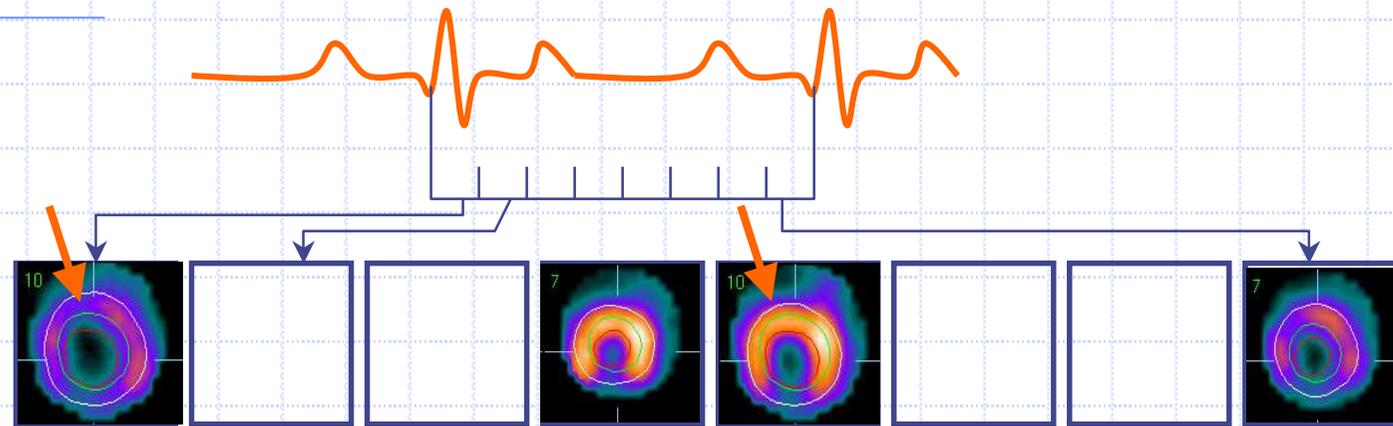


mais FWHM ≈ 15 mm \approx mouvement
qui est donc peu ou pas visible.

Rappel : Effet de volume partiel



EPAISSISSEMENT SYSTOLIQUE

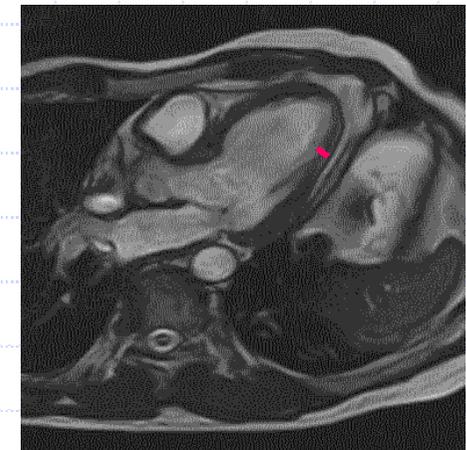


Epaississement systolique ≈ 30 mm

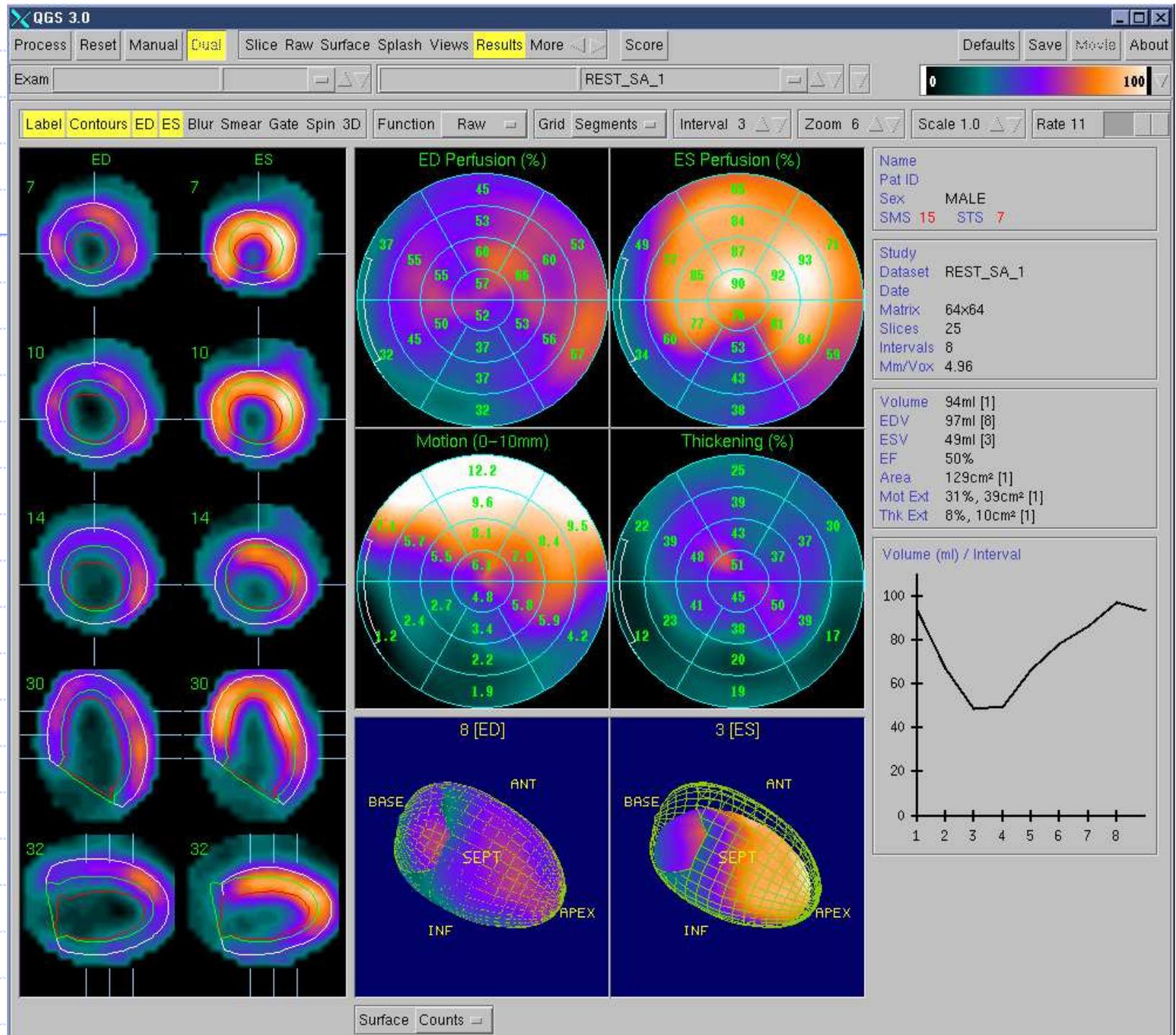
↳ 100 % de l'activité mesurée

Pour une épaisseur diastolique de 15 mm

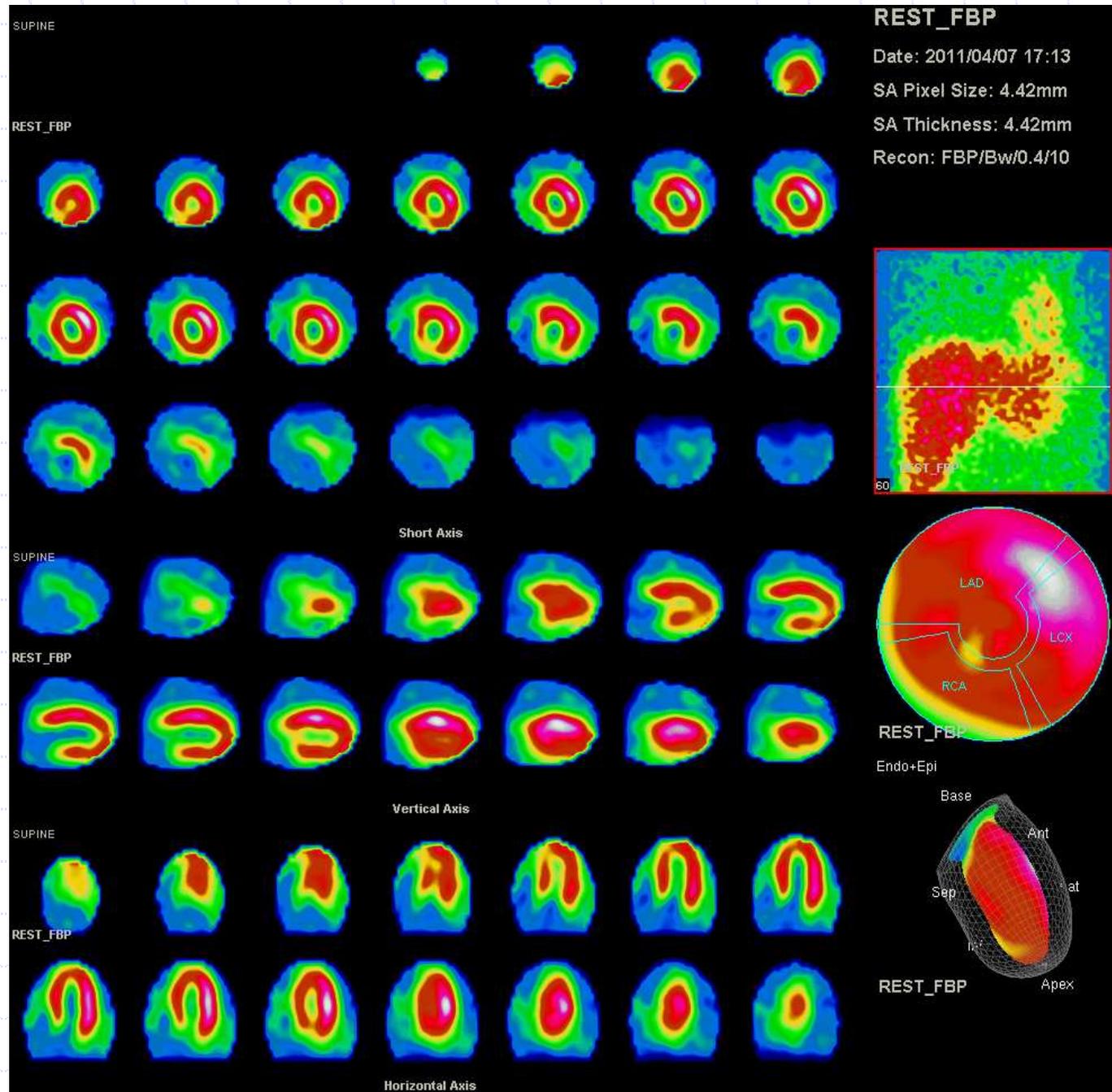
↳ 75 % de l'activité mesurée



QPS[®]
et
QGS[®]

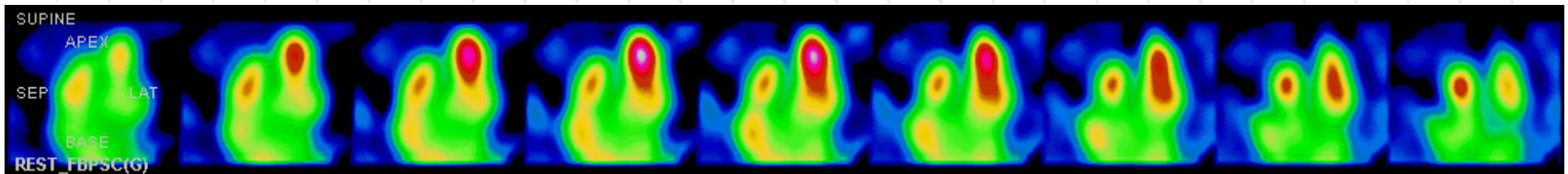
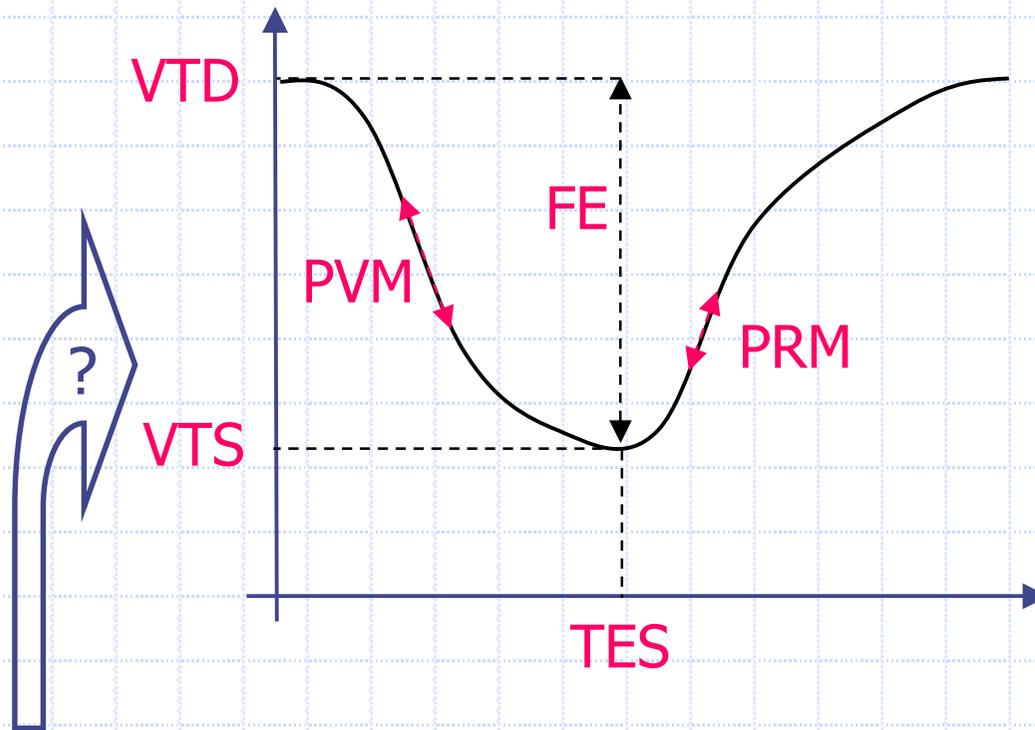
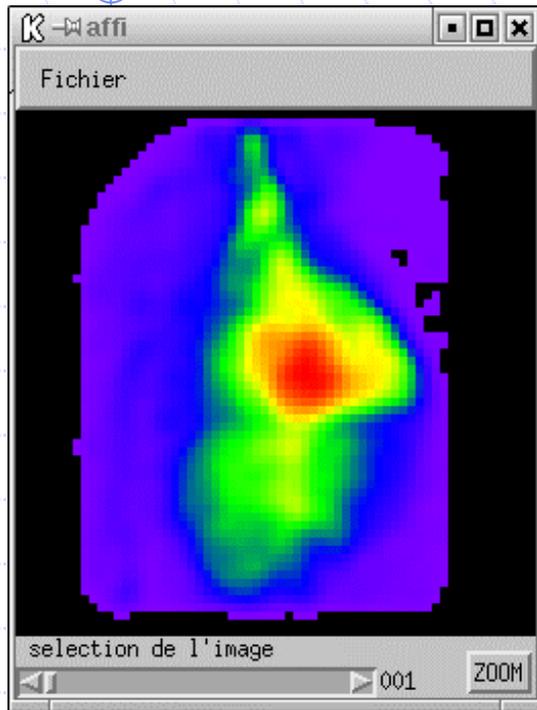


MIBG



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7. Segmentation, analyse et validation (ventriculo)

TOMO-VENTRICULOGRAPHIE



Choix d'un algorithme de segmentation

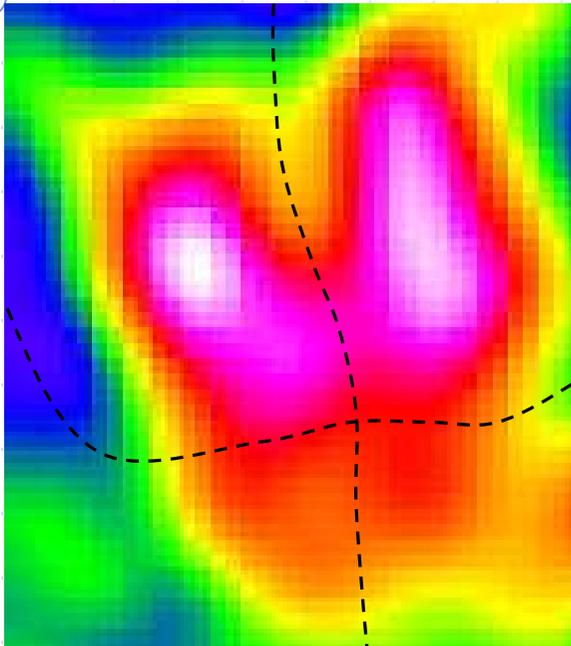
◆ Approche « frontières »

- ◆ Seuillages (simple, d'histogramme, hystérésis...)
- ◆ Dérivatives (gradient, Laplacien)
- ◆ Surfaiques (Hueckel)
- ◆ Morphologiques (gradient, ligne de partage des eaux)
- ◆ Réseau de neurones
- ◆ Champs de Markov
- ◆ Variationnelles

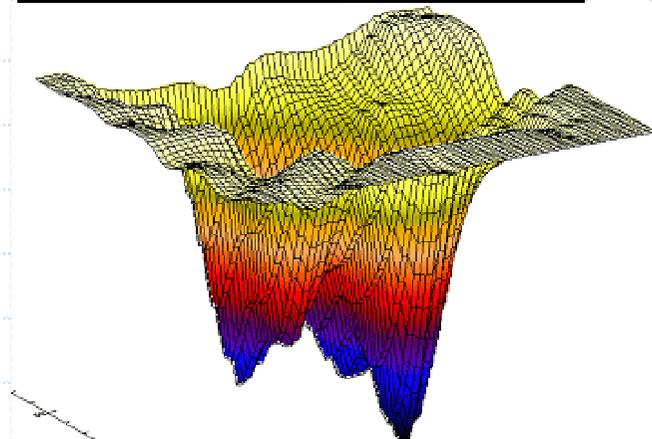
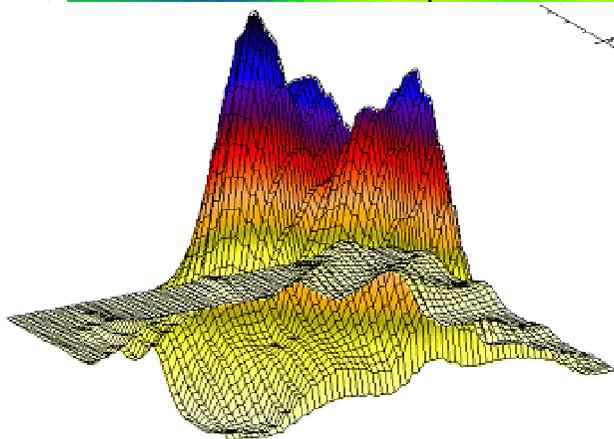
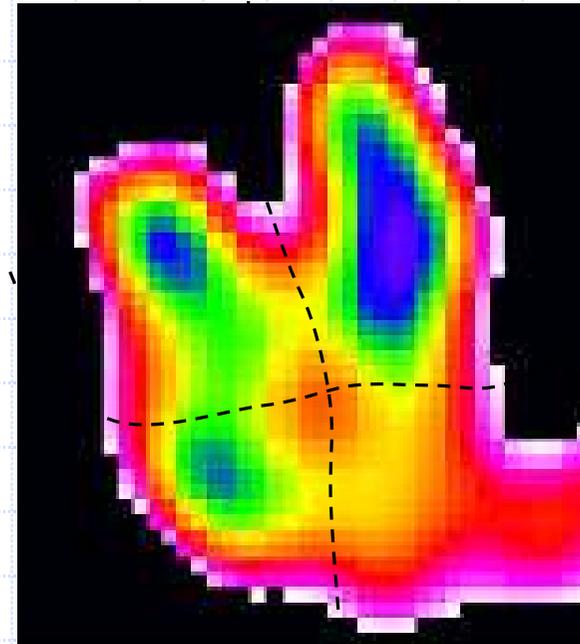
◆ Approche « régions »

- ◆ Techniques Markoviennes,
- ◆ Croissance de régions,
- ◆ Division et fusion de graphes,
- ◆ Etc.

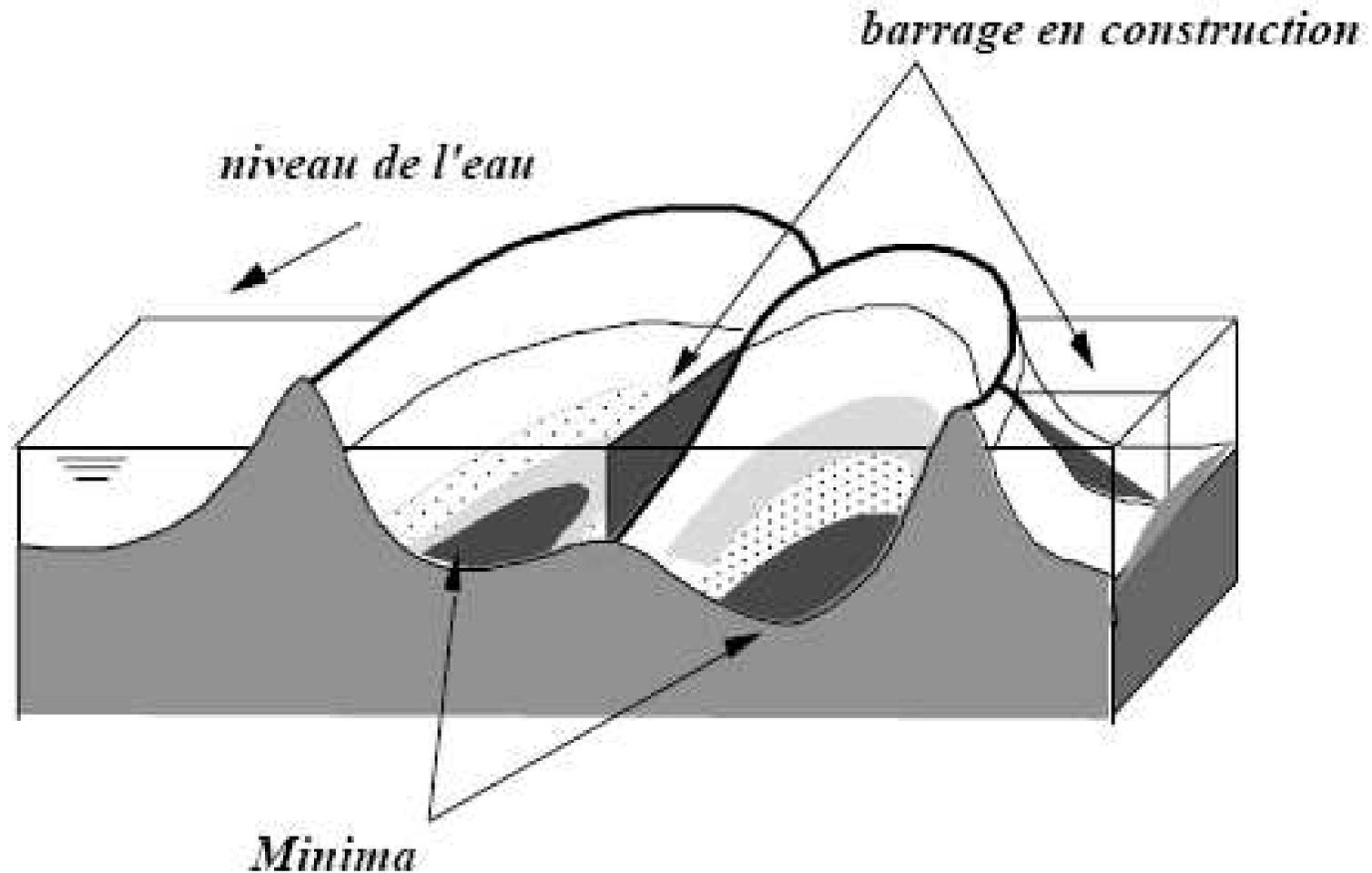
Segmentation



$$NG = NG_{\max} - NG$$

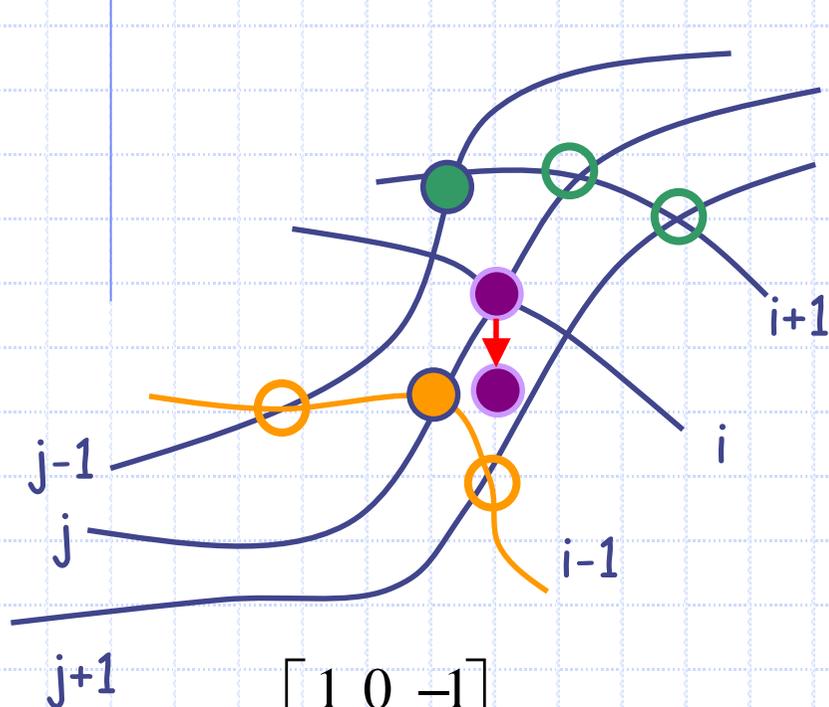


LPE par immersion

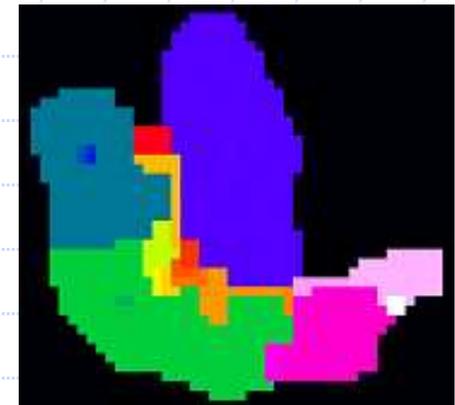
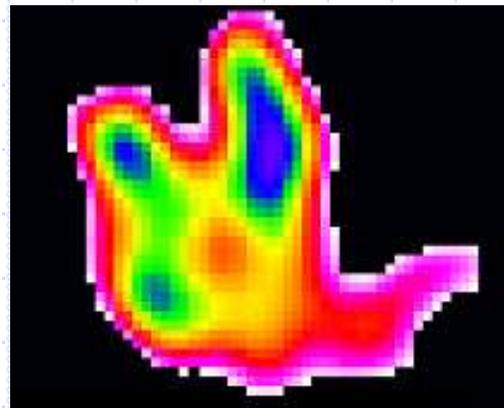


LPE par amincissement homotopique

Si $f_{\max} < f(i, j) \leq f_{\min}$ alors $f(i, j) = f_{\max}$

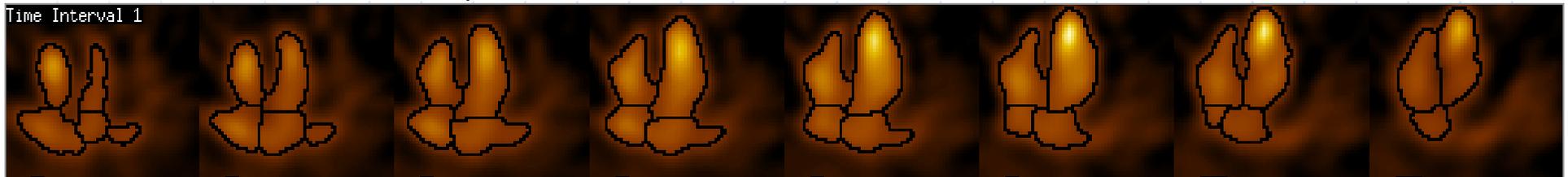
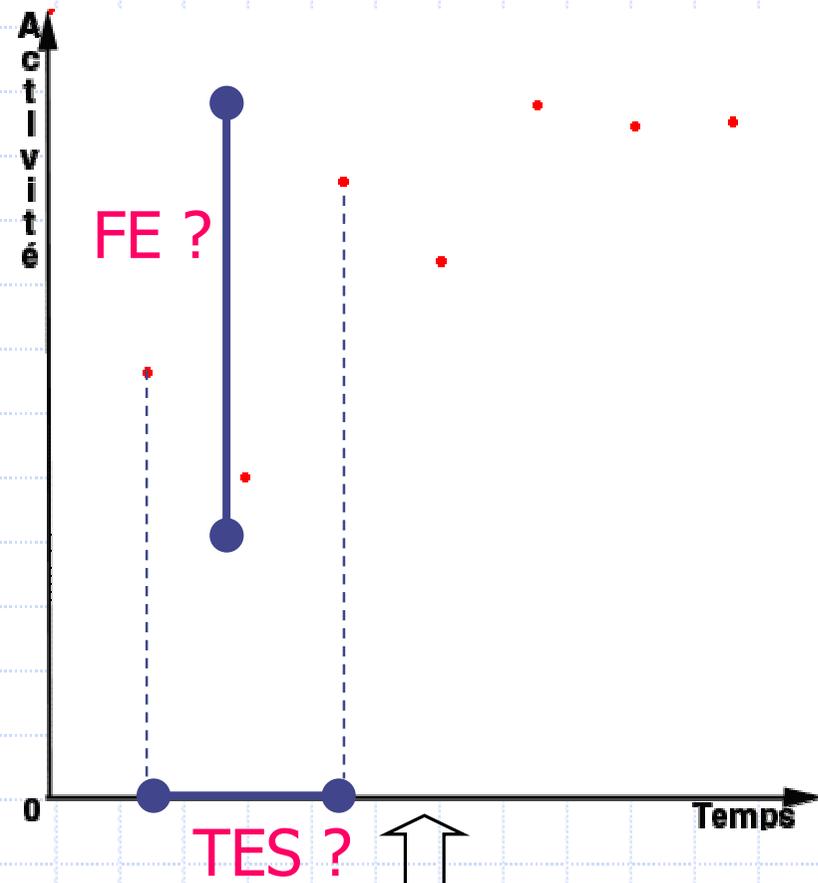
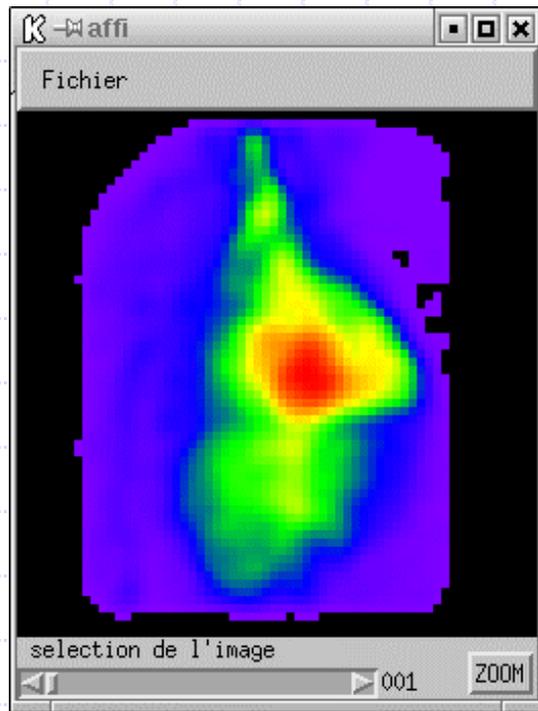


$$L = \begin{bmatrix} 1 & 0 & -1 \\ 1 & 0 & -1 \\ 1 & 0 & -1 \end{bmatrix}$$



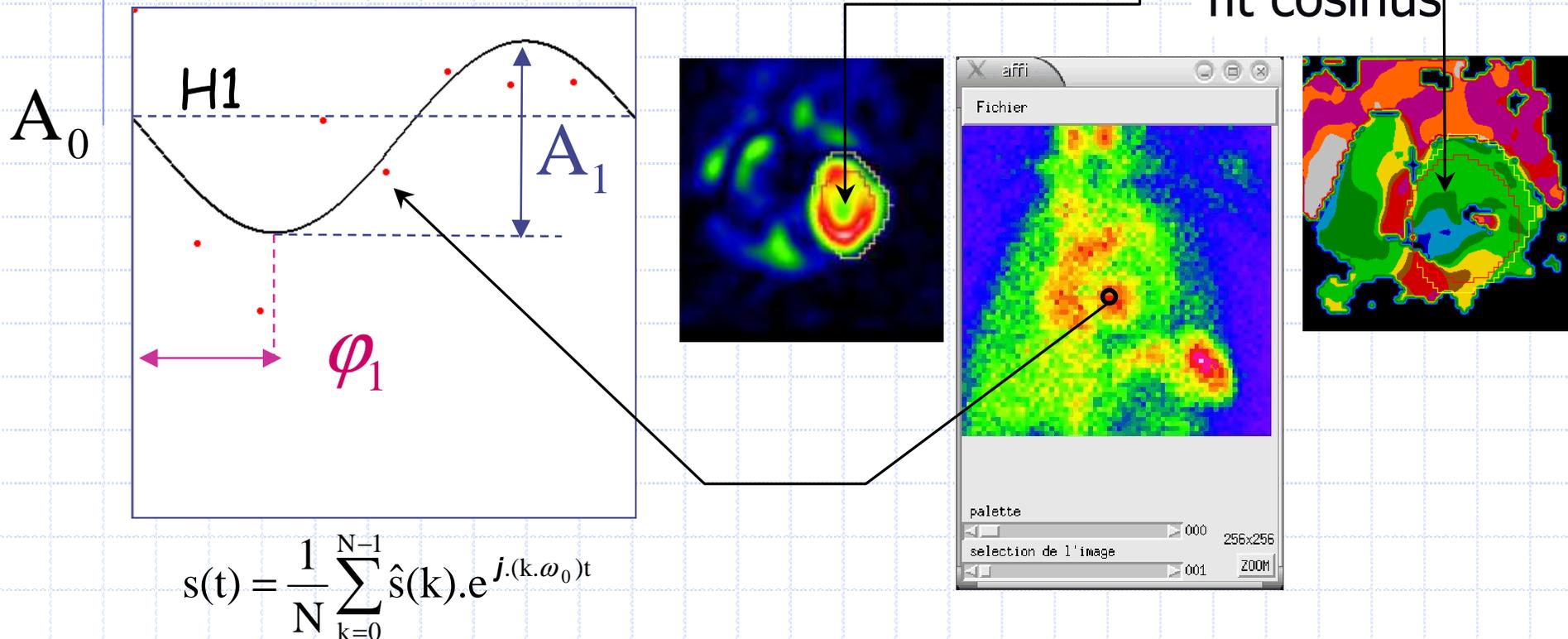
$$L' = \begin{bmatrix} 0 & 1 & 0 \\ 1 & -1 & 1 \\ 0 & 0 & 0 \end{bmatrix}$$

Segmentation



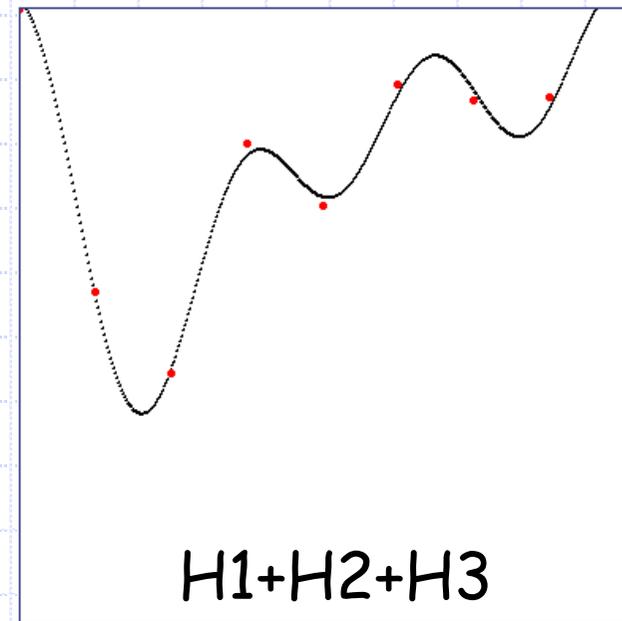
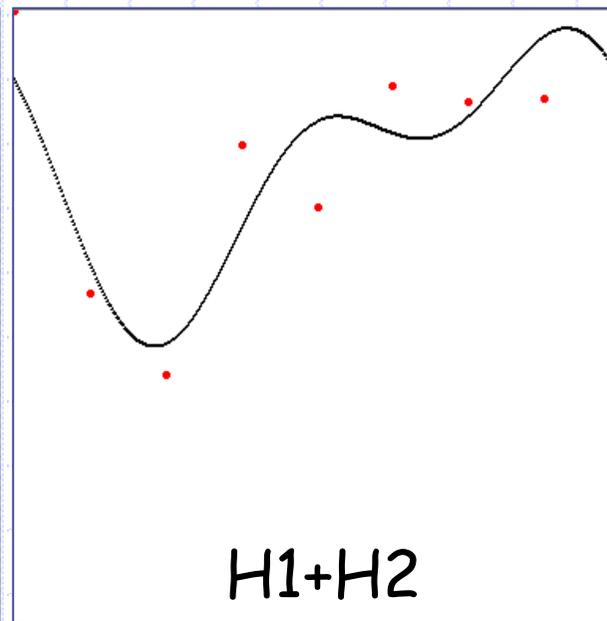
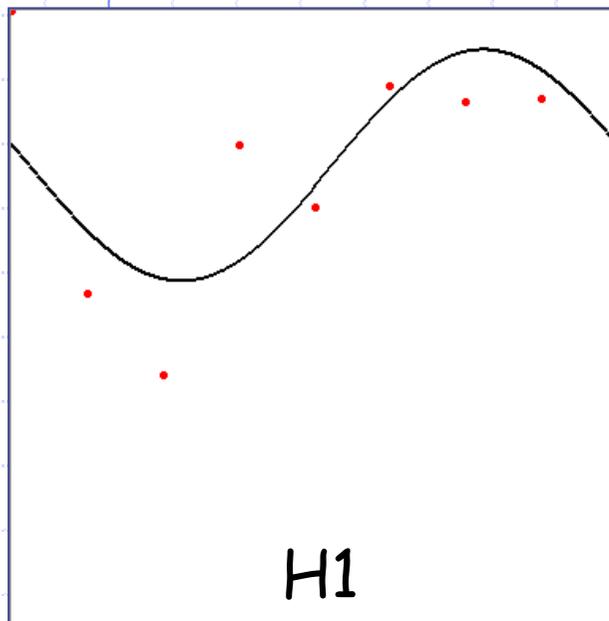
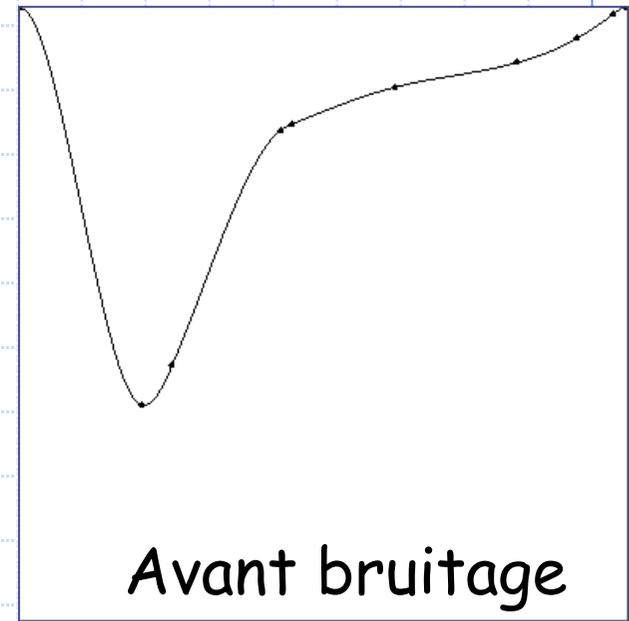
Ajustement mono- harmonique

$$s(t) = \frac{\hat{s}(0)}{N} + \frac{\hat{s}(1)}{N} e^{j.(\omega_0)t} = A_0 + \underbrace{A_1}_{\text{fit cosinus}} e^{j.(\omega_0 t + \varphi_1)}$$

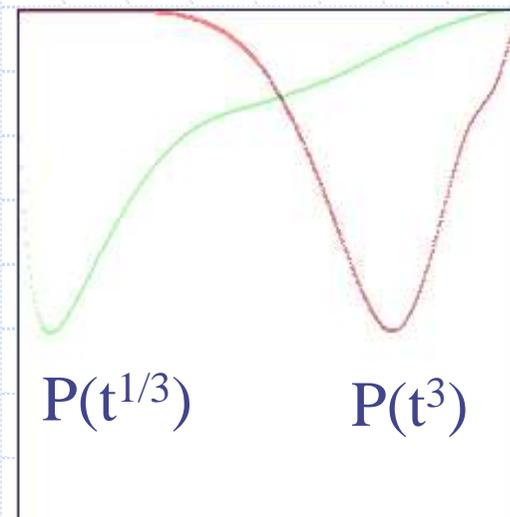
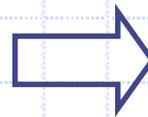
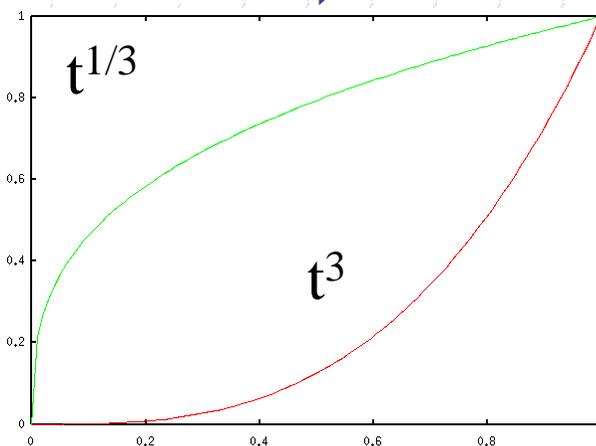
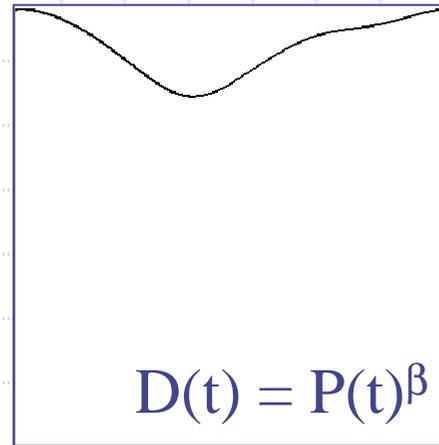
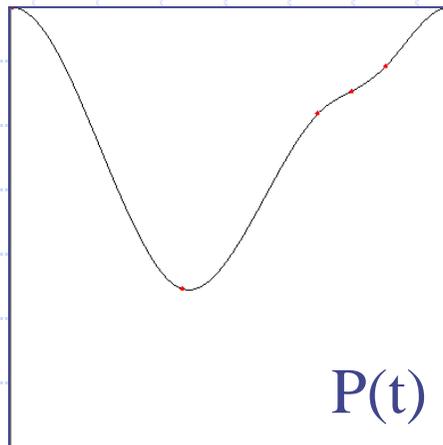


Ajustement multi-harmonique

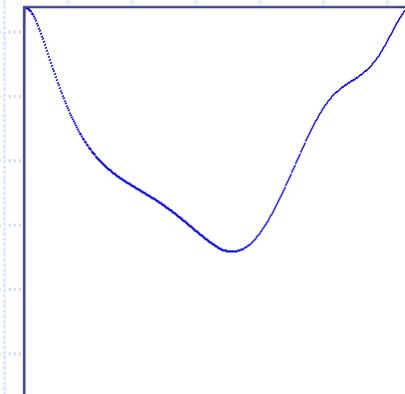
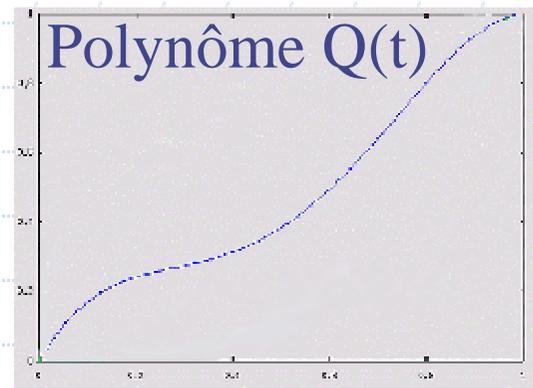
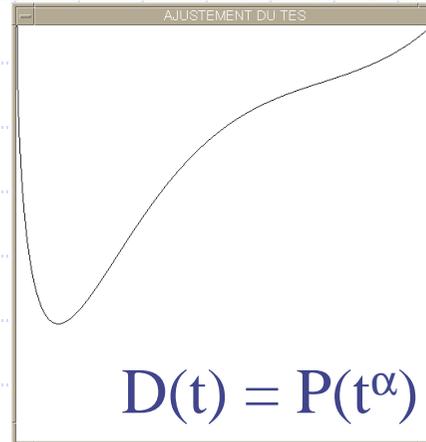
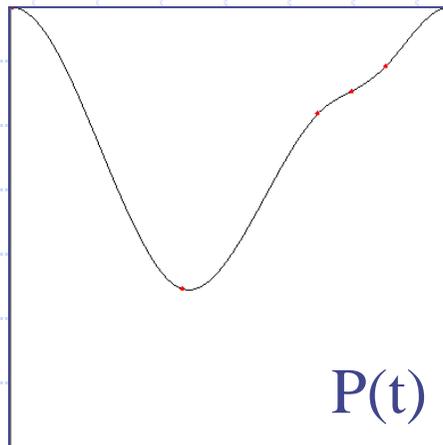
$$s(t) = \frac{1}{N} \sum_{k=0}^{H-1} \hat{s}(k) \cdot e^{j \cdot (k \omega_0) t}$$



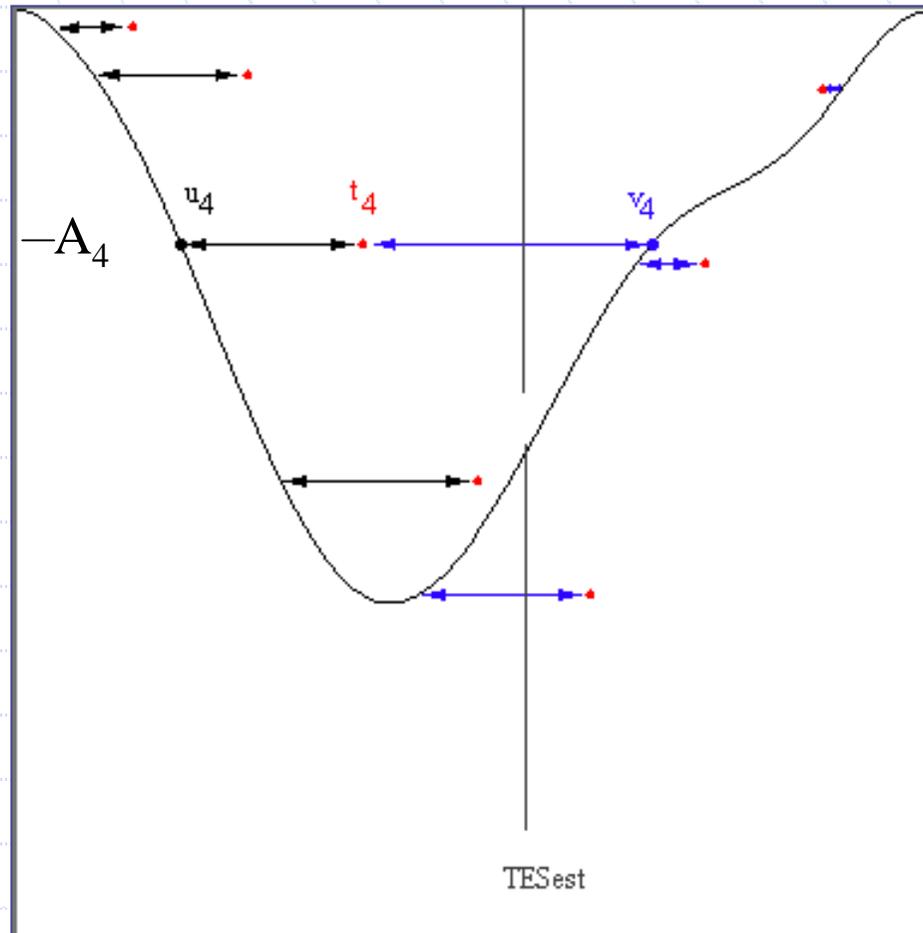
Déformation d'une CTA de référence



Déformation d'une CTA de référence



Détermination de $Q(t)$

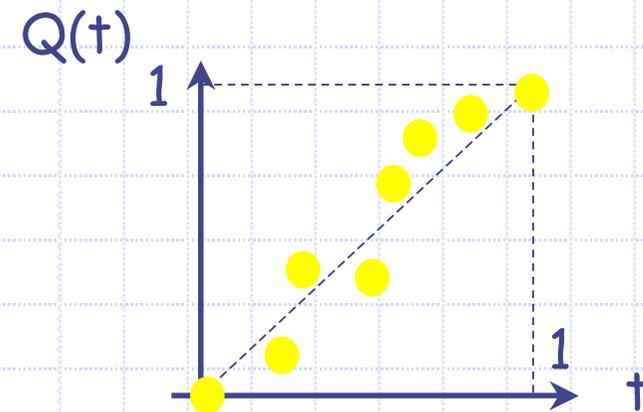


Acquisition bruitée (t_4, A_4)

$$A_4 = P(u_4)$$

$$A_4 = D(t_4) = P[Q(t_4)]$$

$$Q(t_4) = u_4$$

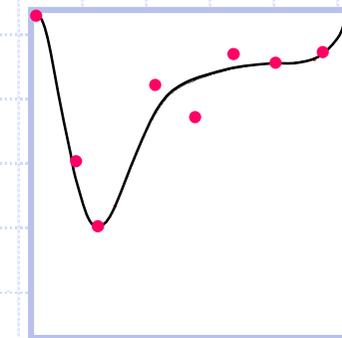
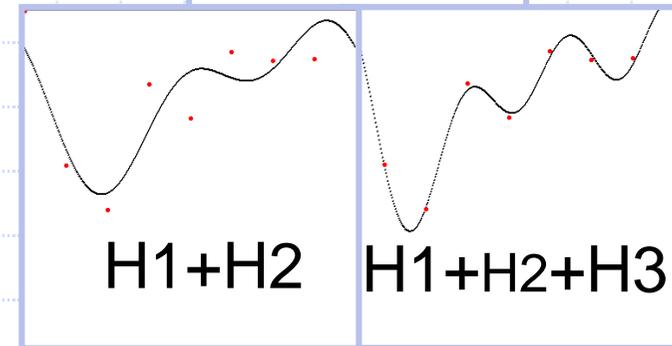
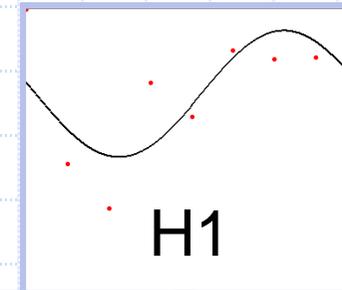
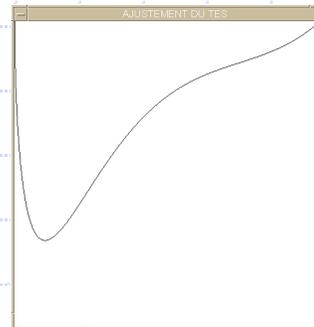
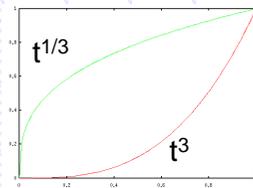
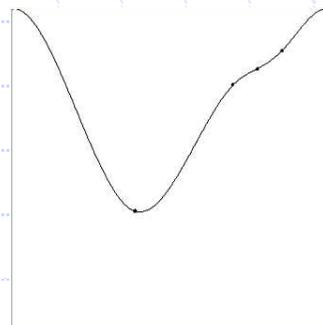


Validation

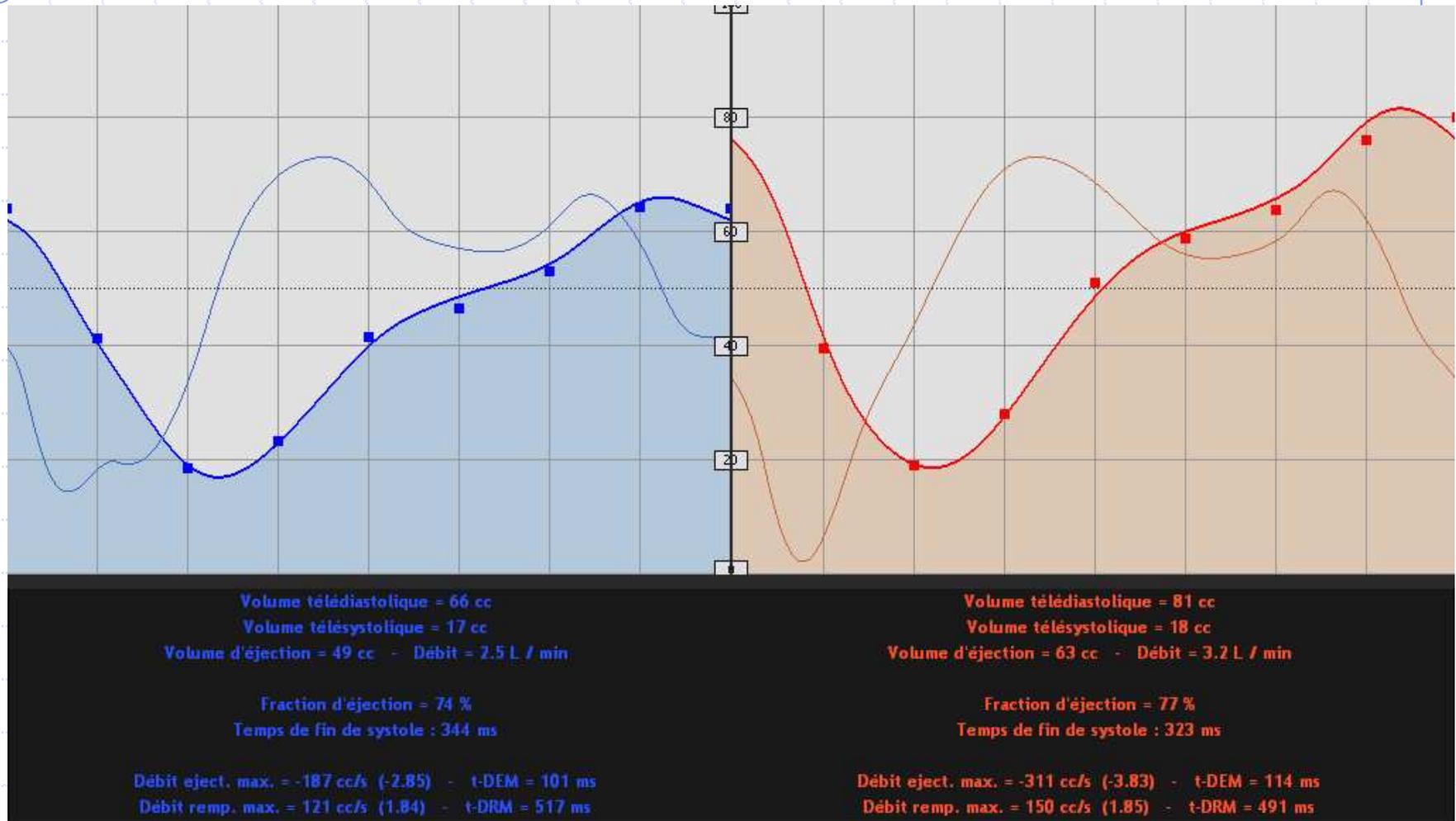
◆ Filtrage linéaire de la CTA

- H1 : imprécis
- Multi-harmoniques : bruité

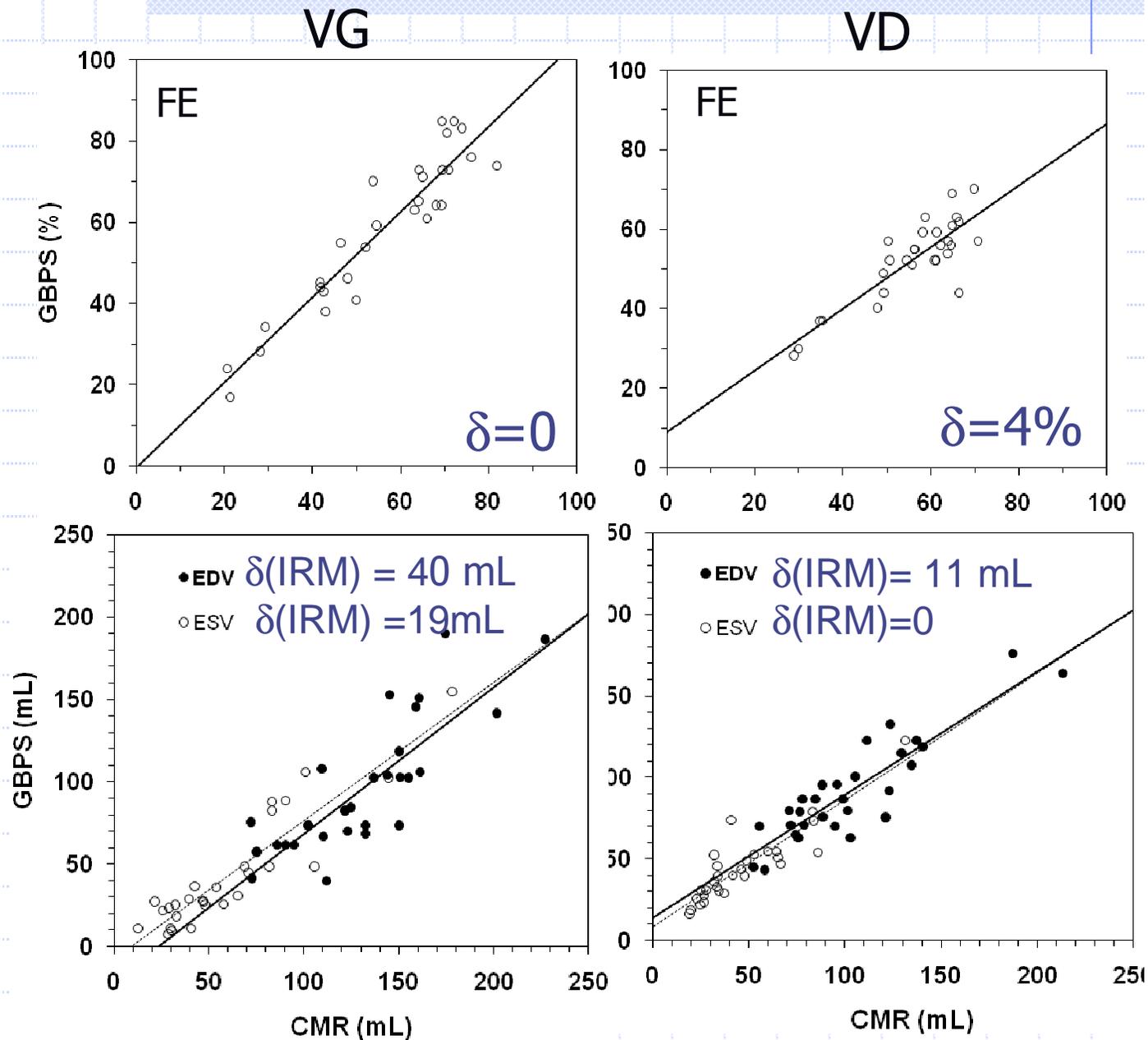
◆ Modélisation de la CTA



Analyse de la fonction systolique



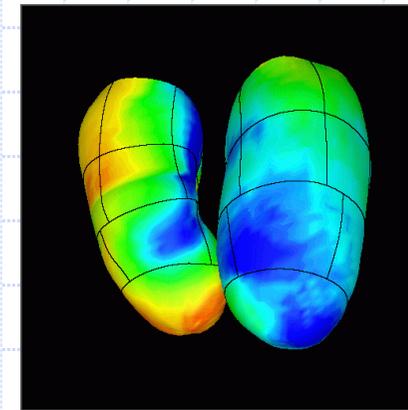
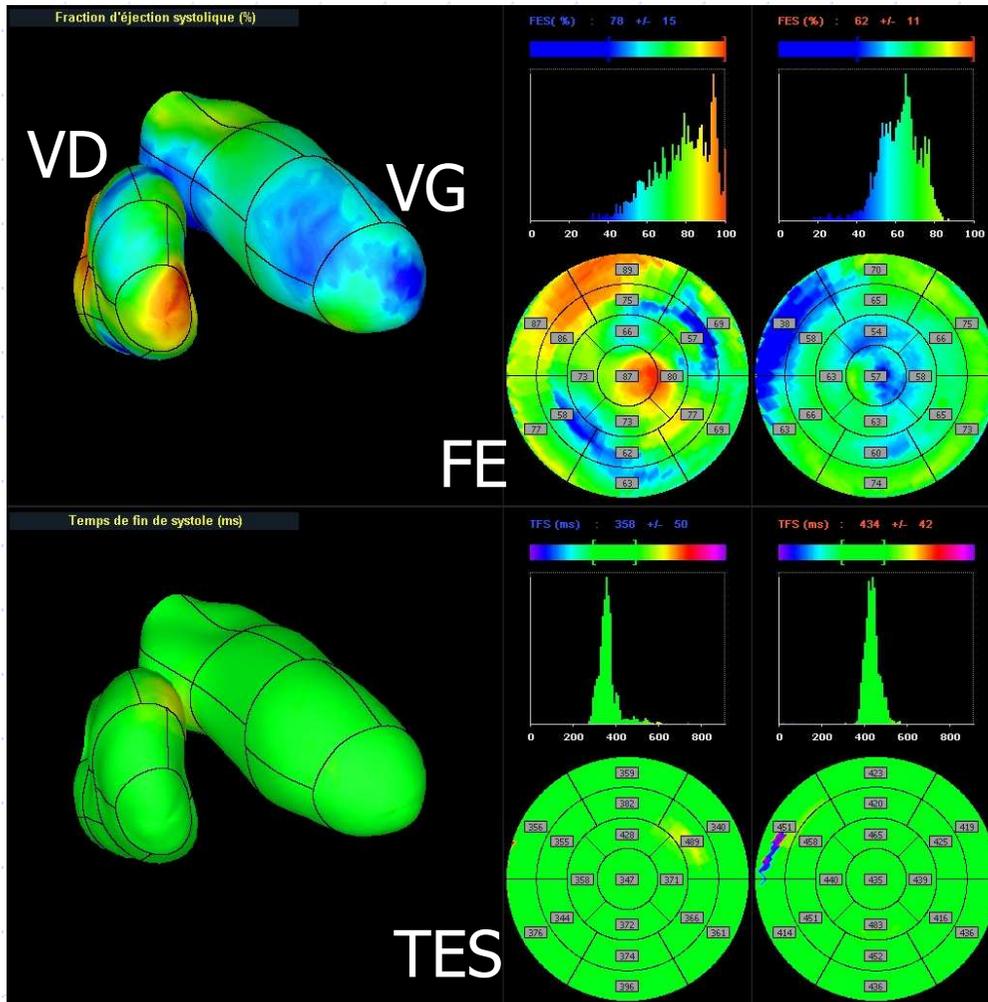
TVI versus IRM



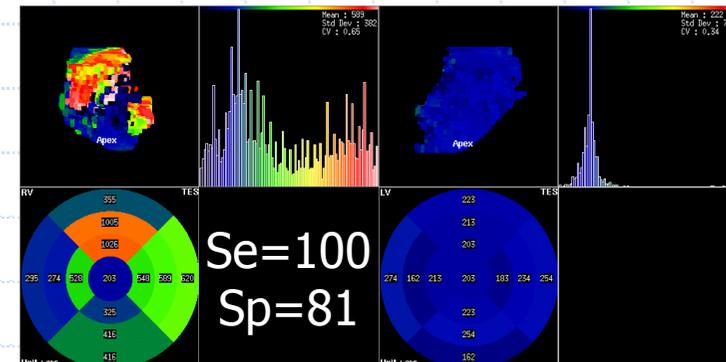
VES(G-D): 9 ± 14 (GBPS) versus 18 ± 13 (IRM)

L. Sibille et D. Mariano-Goulart, An. Nucl. Med 2010.

ANALYSE 3D DE CTA LOCALES



Parameter	Control subjects	Patients with localized ARVD	Patients with diffuse ARVD
EF (%)	63 ± 7	60 ± 5	33 ± 12 [†]
EDV (mL)	101 ± 17	107 ± 20	180 ± 81 [†]
ESV (mL)	37 ± 8	43 ± 10	124 ± 70 [†]
σ-EF (%)	13 ± 3	14 ± 2	11 ± 3
σ-TES (ms)	63 ± 24	167 ± 64 [†]	277 ± 106 [†]

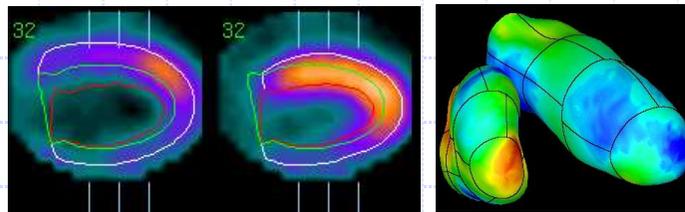
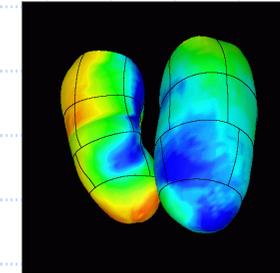


S=80 ms

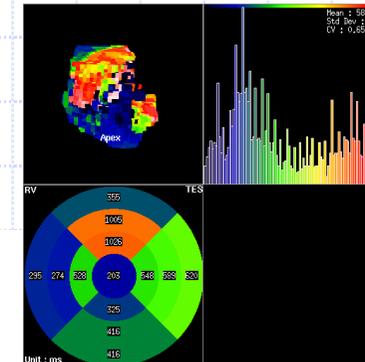
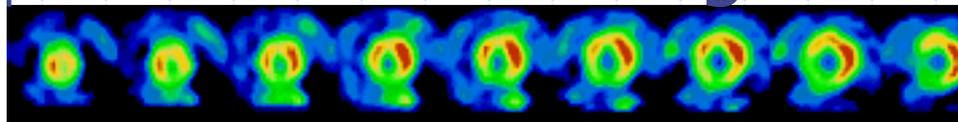
Se=100
Sp=81

Conclusion : patients à risque

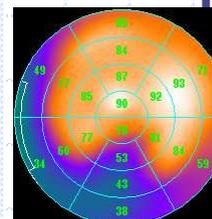
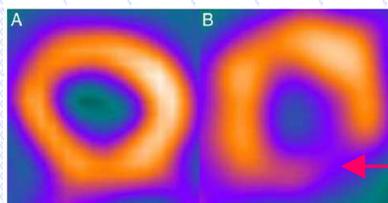
- ◆ Fraction d'éjection (G,D) ↓
 - ◆ $FE(VG) < 35\% \Rightarrow$ Défibrillateur implantable
- ◆ Volumes ventriculaires (G,D) ↑

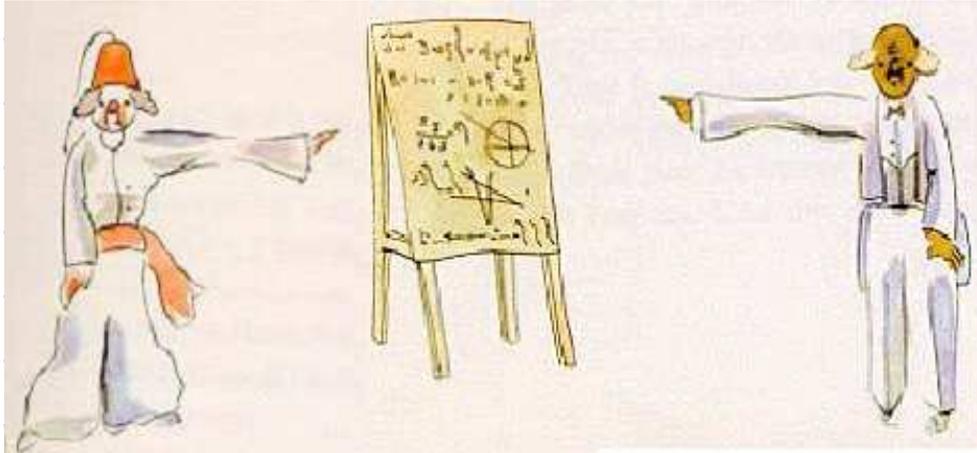


- ◆ Dépolarisation hétérogène



- ◆ Innervation sympathique ↓





The Mathematics of
Computerized Tomography.
F. Natterer. 2001. SIAM.

Introduction au traitement
numérique des images.
D. Mariano-Goulart
Encyclopédie Médico-chirurgicale,
35-100-A-10, 2009.

Reconstruction tomographique en
imagerie médicale.
D. Mariano-Goulart
Encyclopédie Médico-chirurgicale,
35-105-A-10, 2009.

<http://scinti.etud.univ-montp1.fr>

Merci de votre attention...

d-mariano_goulart@chu-montpellier.fr

Résumé

Ce cours vise à illustrer le rôle qu'un spécialiste en traitement et analyse d'images peut jouer au sein de services hospitaliers ou de sociétés commerciales liées à l'imagerie médicale. Nous partirons d'un problème clinique spécifique qui se pose aux cardiologues (évaluation pronostique des insuffisances cardiaques) pour envisager le choix d'une modalité d'imagerie médicale et d'un protocole d'acquisition d'images. Nous évoquerons ensuite comment le signal peut être déconvolué de la réponse impulsionnelle de l'appareil d'imagerie, puis bénéficier d'une reconstruction tomographique. Les problématiques liées à la segmentation des données d'imagerie puis à leur analyse sera enfin présentée en concluant sur les paramètres cliniques extraits et utilisés en routine hospitalière. Les solutions logicielles qui seront présentées au fil de l'exposé ont toutes été mises au point dans les vingt dernières années et constituent donc une image assez représentative de l'apport des méthodes de traitement du signal à l'amélioration de la prise en charge médicale de certaines pathologies.