TRAITEMENT D'IMAGES MEDICALES

Illustration du rôle d'un ingénieur au sein d'une structure hospitalière

Module « Bioingénierie, 8° semestre, 2011

Ecole Centrale de Marseille.

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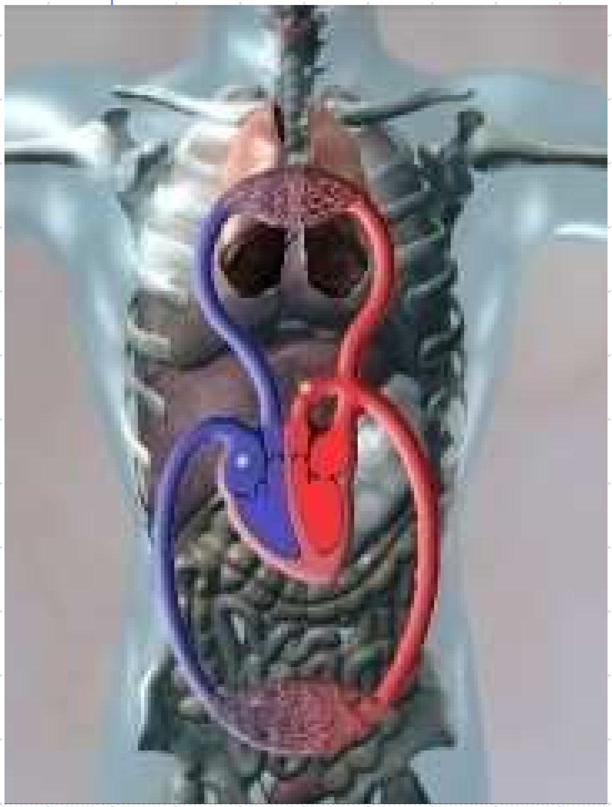
http:\\scinti.etud.univ-montp1.fr

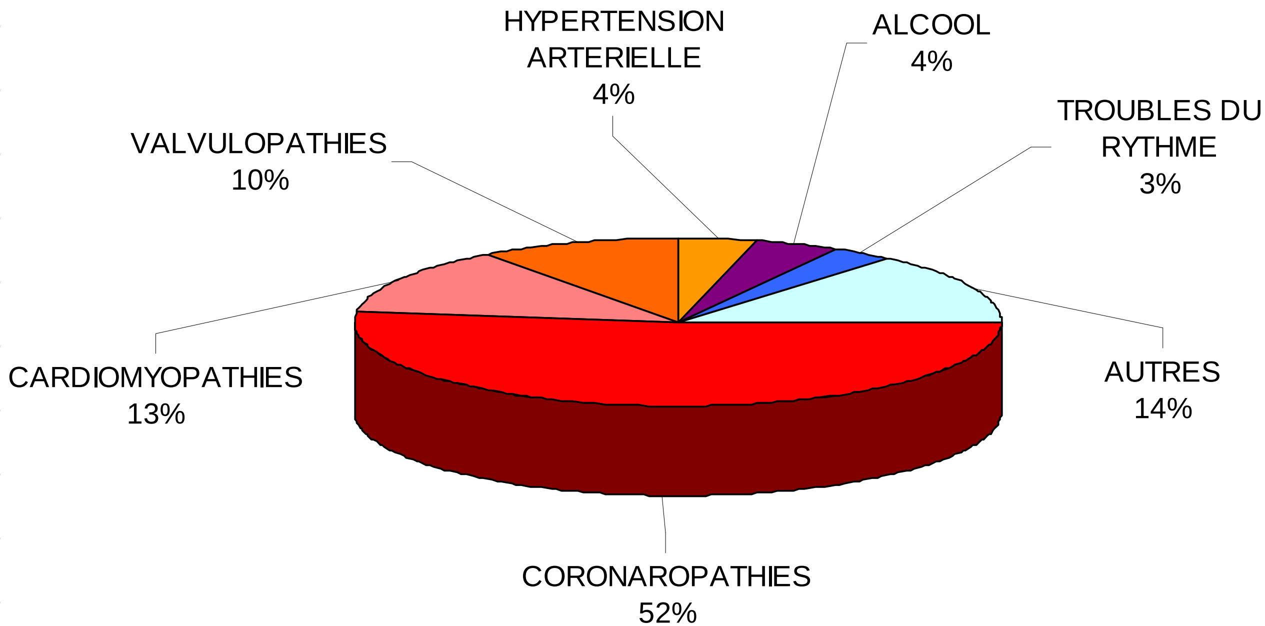
PLAN

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



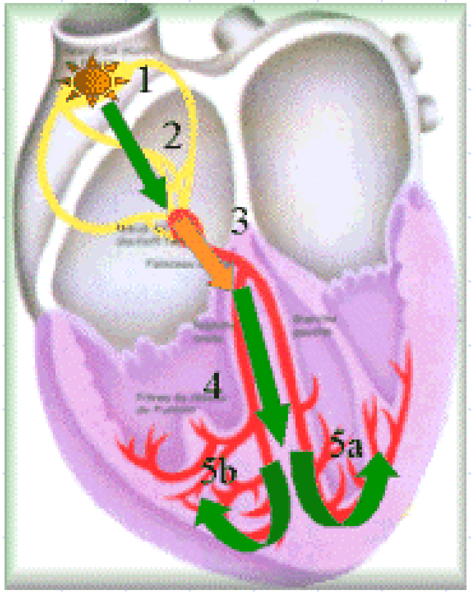


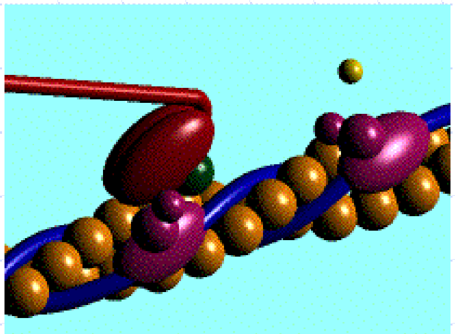
Prévalence = 1-3 % (1 million en France)

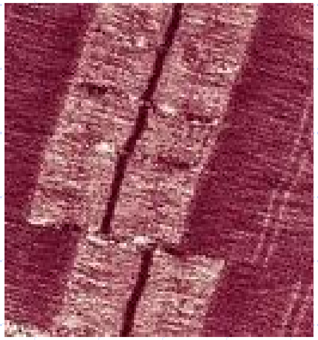
A 5 ans : 25 (H) à 40 %(F) de décès
Environ ½ morts subites

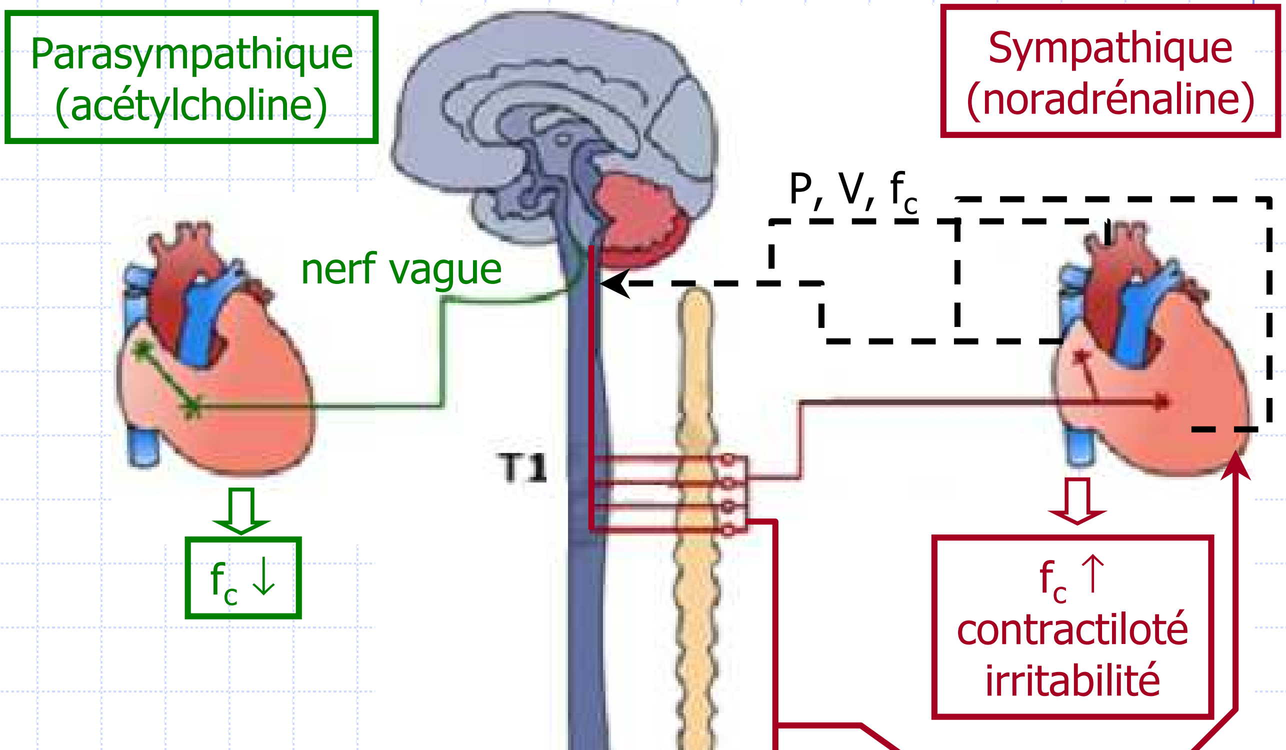
PHYSIOLOGIE CARDIAQUE

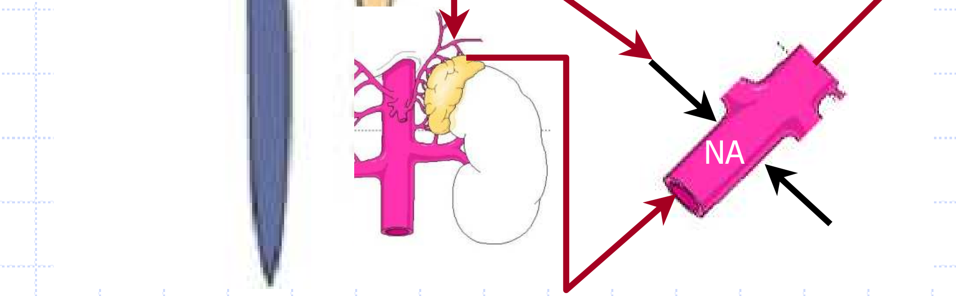
Automatisme cardiaque



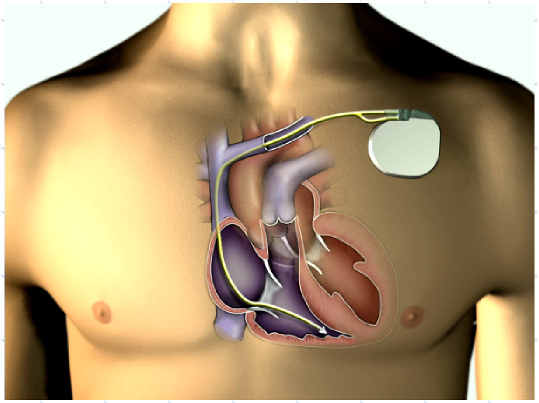








PROBLEME CLINIQUE



- Mesures hygiéno-diététiques
- Médicaments (Diu., IEC, βB)
- Défibrillateur implantable si $FE = \frac{VTD - VTS}{VTD} < 35\%$
  - mais décès avec FE > 40%
  - 35 % des défibrillateurs implantés sont sollicités / 3 ans.
- Meilleure identification des patients à risque de mort subite par imagerie ?
  - Quelle fonction cibler (pompe, innervation, perfusion) ?
  - Quelle modalité d'imagerie ?
  - Quels traitements d'image ?

AJ Moss AJ et al. Long-term clinical course of patients after termination of ventricular tachyarrhythmia by an implanted defibrillator.Circulation 2004;110:3760 –5

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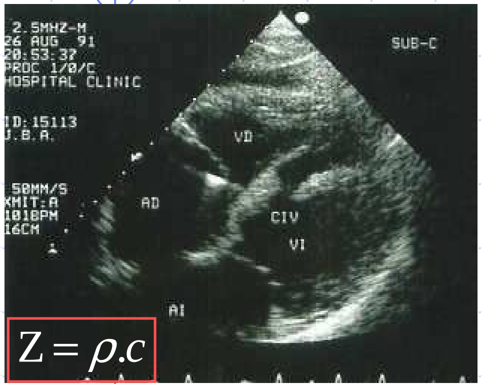
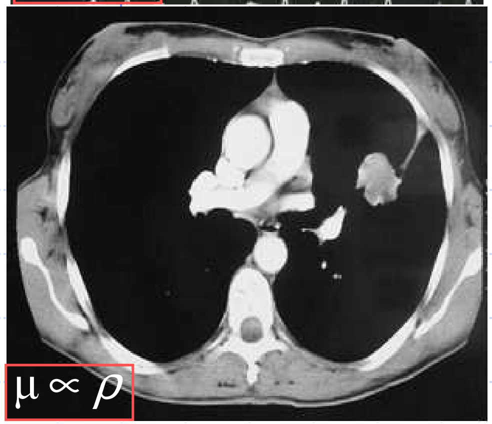
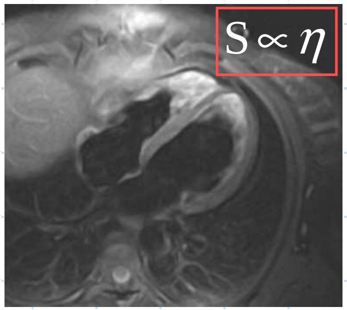
CHOIX D'UNE MODALITE D'IMAGERIE

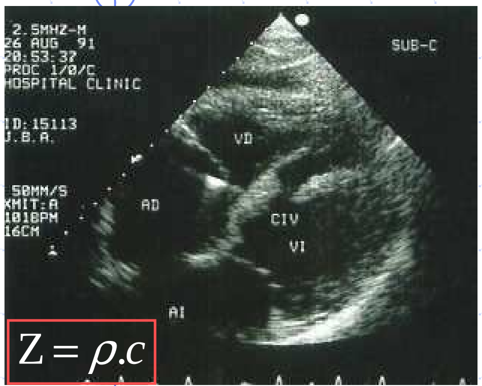
ANATOMIQUE



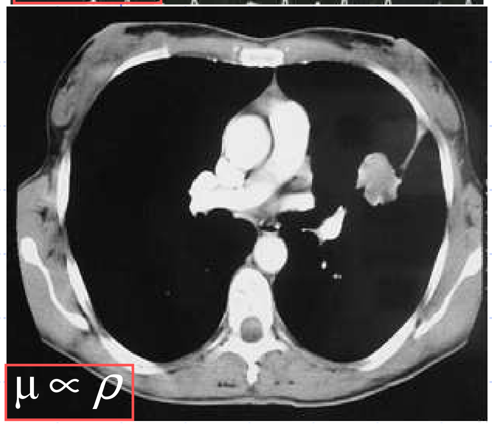




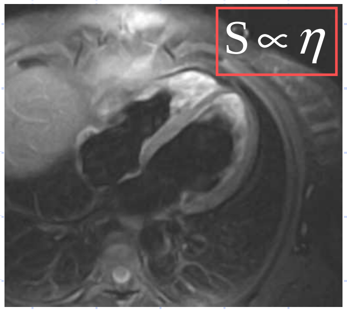
CHOIX D'UNE MODALITE D'IMAGERIE



$Z = \rho . c$



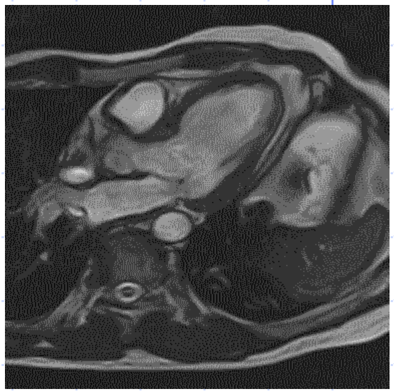
$\mu \propto \rho$



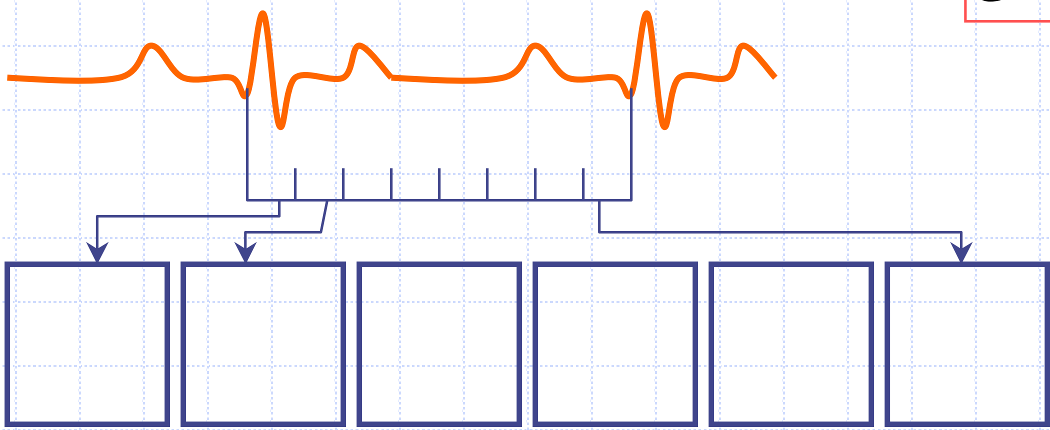
$S \propto \eta$

ANATOMIQUE

FONCTIONNELLE

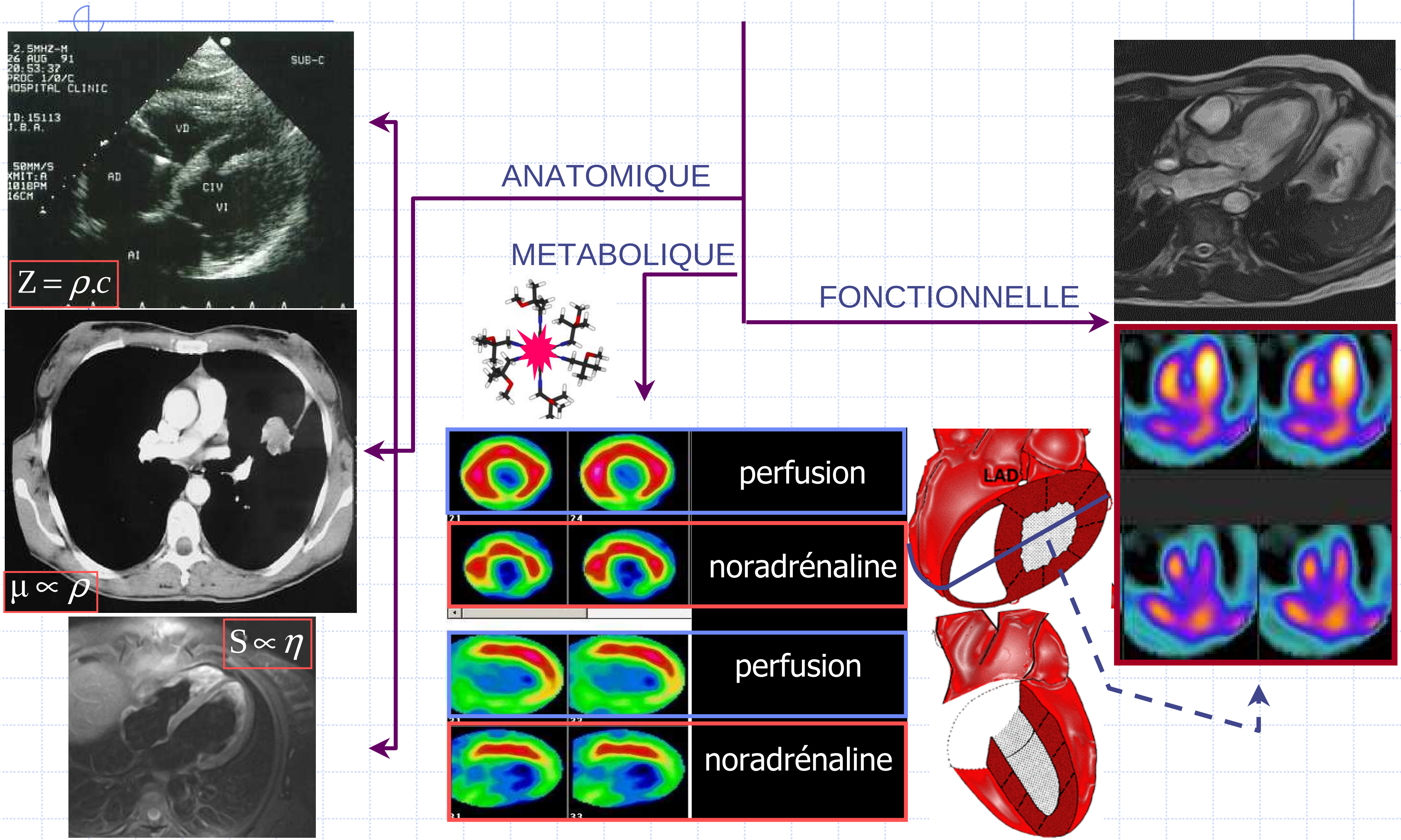


$C = f(t)$



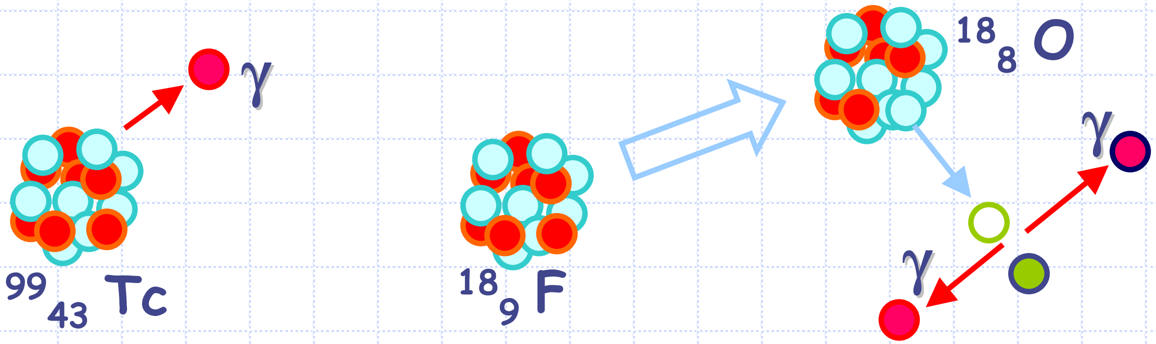
Synchronisation à l'électrocardiogramme

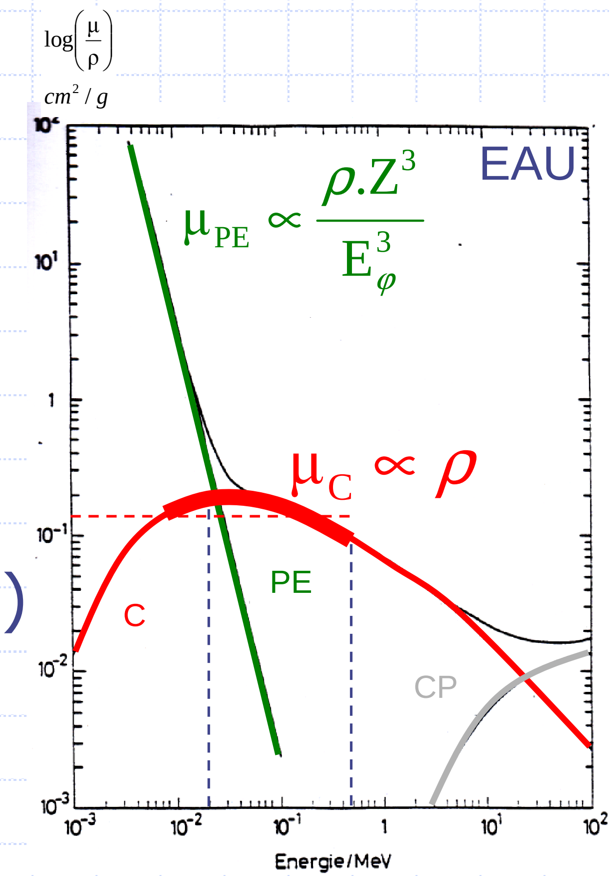
CHOIX D'UNE MODALITE D'IMAGERIE



CHOIX D'UN MARQUEUR RADIOACTIF

- Photons pénétrants et peu diffusés
  - Émetteurs gamma ou bêta plus (ionisants)

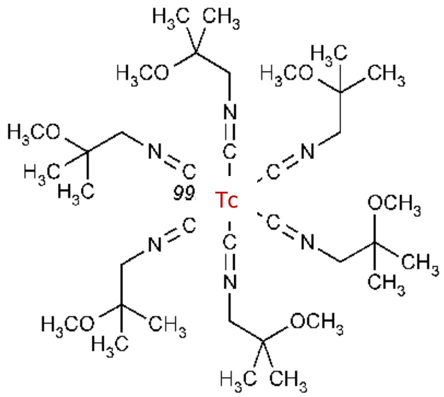


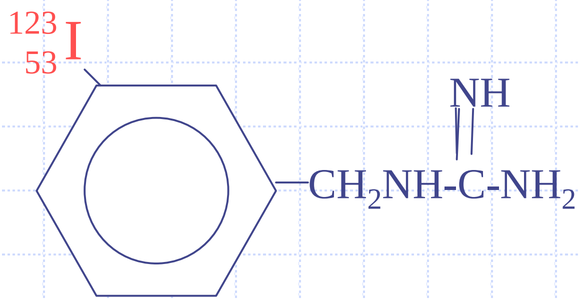


- énergies détectables
  - γ (140 keV $^{99}_{43}Tc$ ou 159 keV $^{123}_{53}I$)
  - $\beta^+ \rightarrow 2\gamma$ 511 keV ($^{18}_{8}F$)

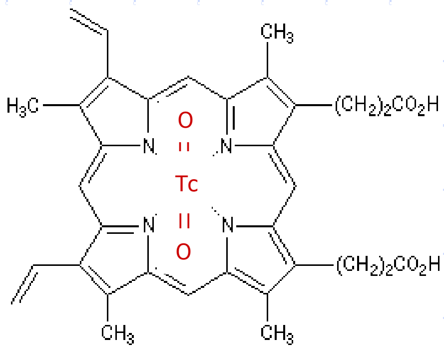
CHOIX D’UN RADIOTRACEUR

- Traçage possible d’un métabolisme
  - Marquage de cations lipophiles (MIBI) au $^{99}_{43}\mathrm{Tc}$

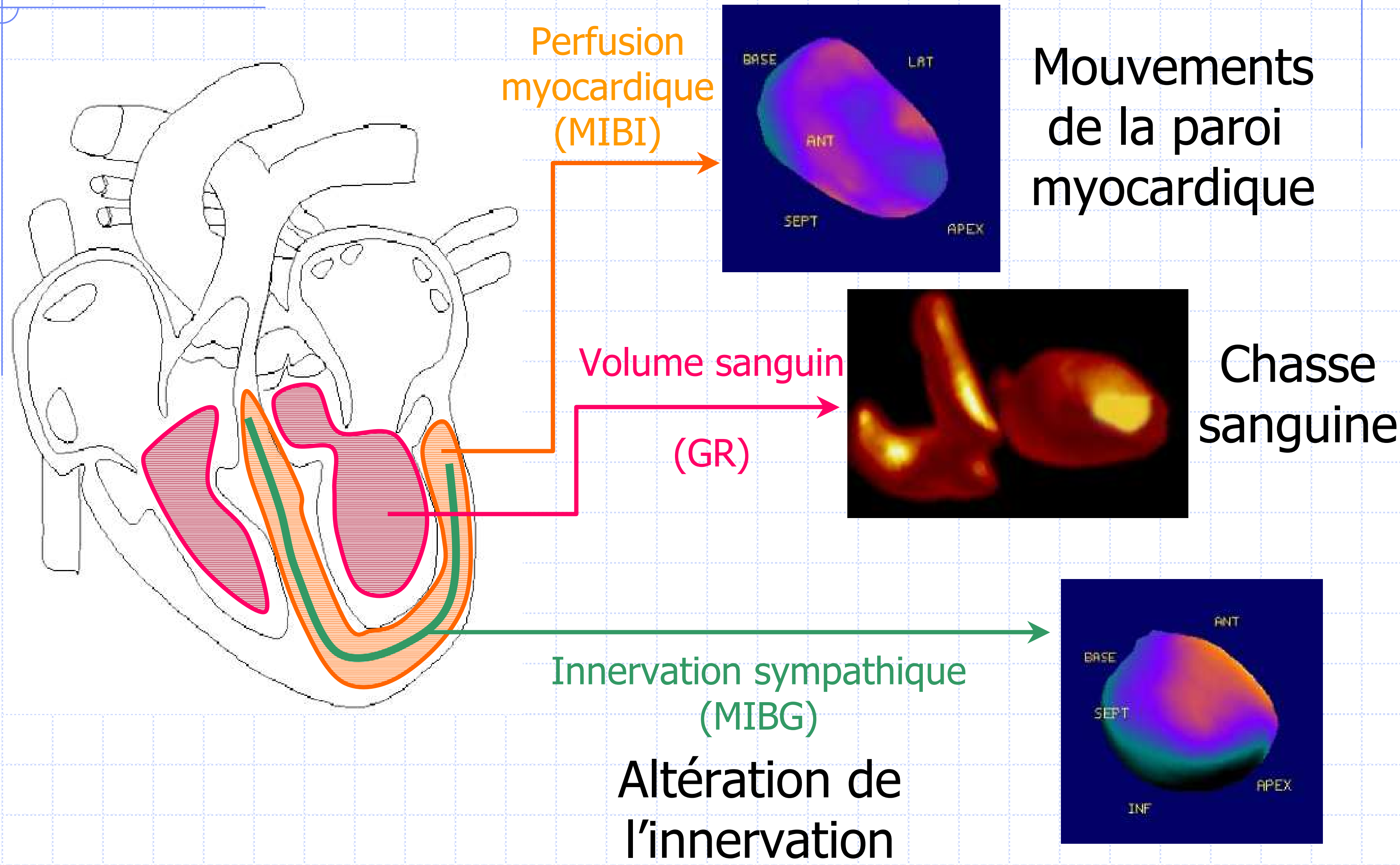




  - Marquage de la MIBG à l’$^{123}_{53}\mathrm{I}$
  - Marquage des globules rouges au $^{99}_{43}\mathrm{Tc}$
    - Chimie de complexation avec le noyau tétrapyrolique de l’hème



CHOIX D'UNE MODALITE D'IMAGERIE



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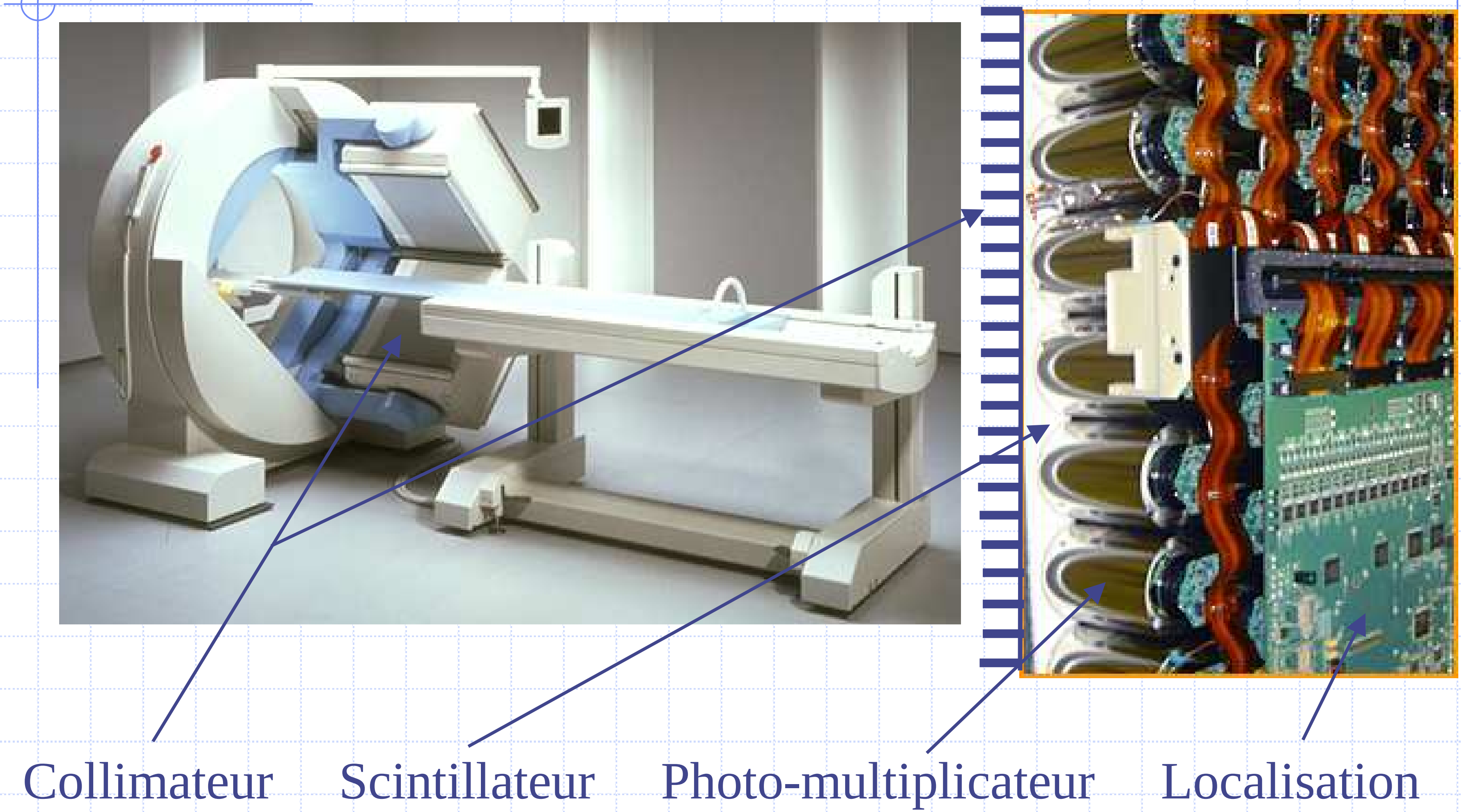
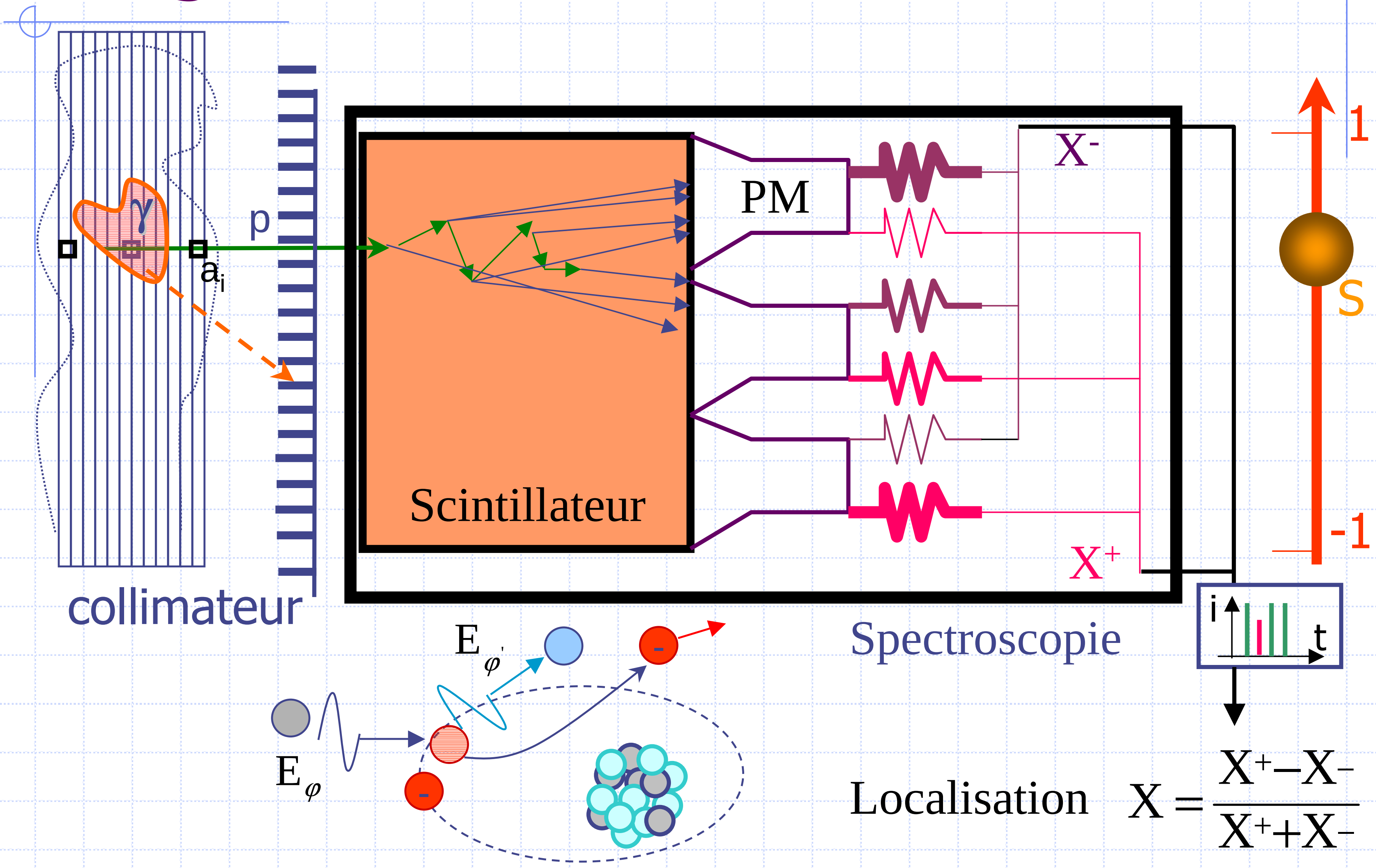
Single Photon Emission CT



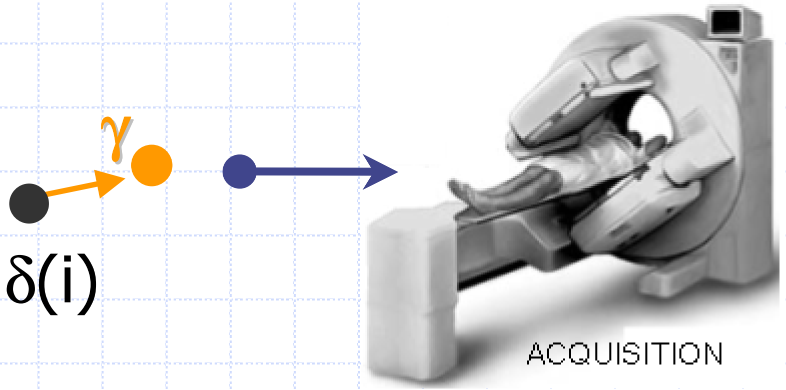
Single Photon Emission CT

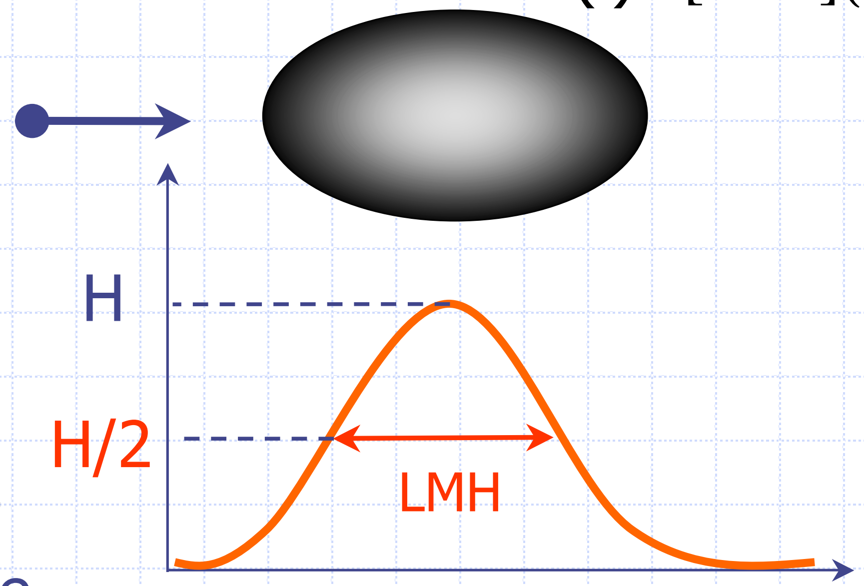


Localisation $X = \dfrac{X^+ - X^-}{X^+ + X^-}$

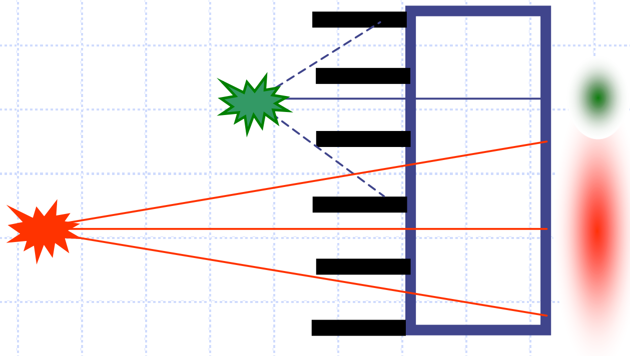
Largeur à mi-hauteur (FWHM)

$h(i)=[h\star\delta](i)$

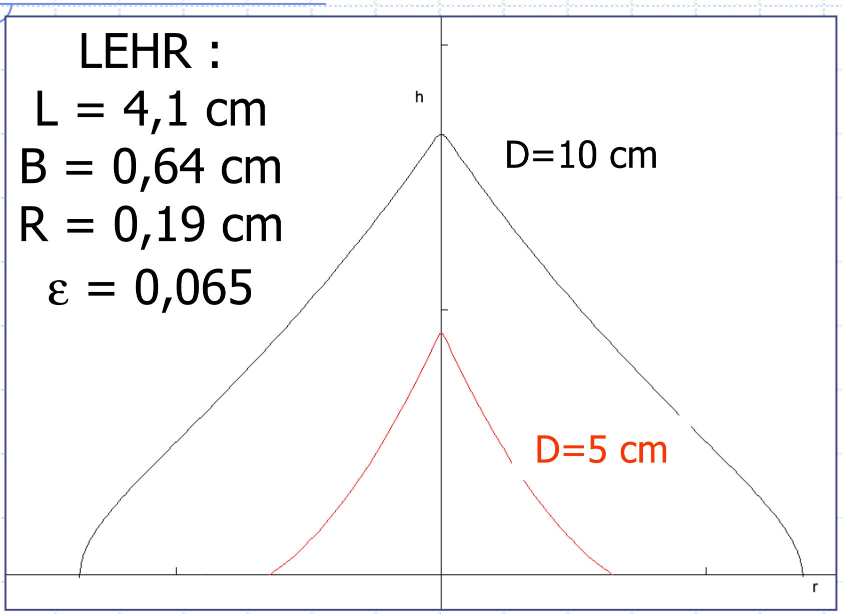


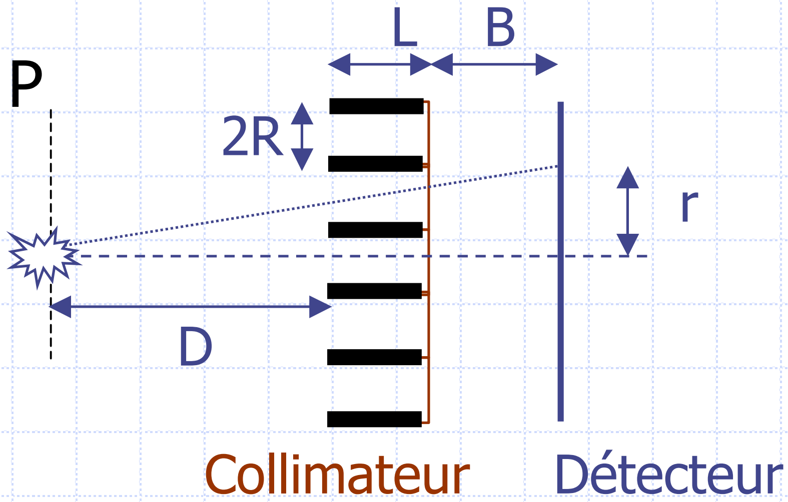


- Réponse intrinsèque h_i
  - LMH ≤ 4 mm, ± invariante
- Réponse du collimateur h_c
  - La LMH dépend de la distance source-détecteur



Réponse d'un collimateur





$$\lambda = \frac{L}{R.(L+D+B)}$$

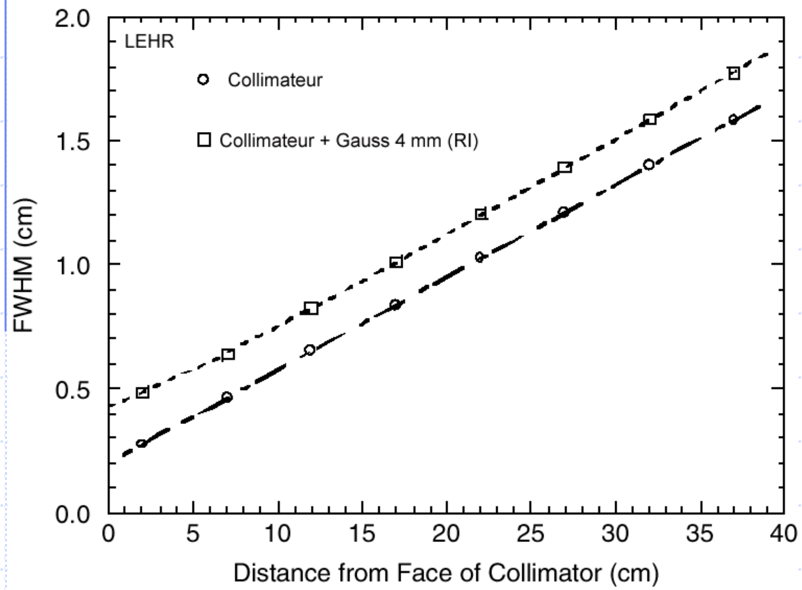
$$h(r) = \frac{\varepsilon}{\pi}\left[2.\arccos\left(\frac{\lambda.r}{2}\right) - \lambda.r.\sqrt{1-\left(\frac{\lambda.r}{2}\right)^2}\right]$$

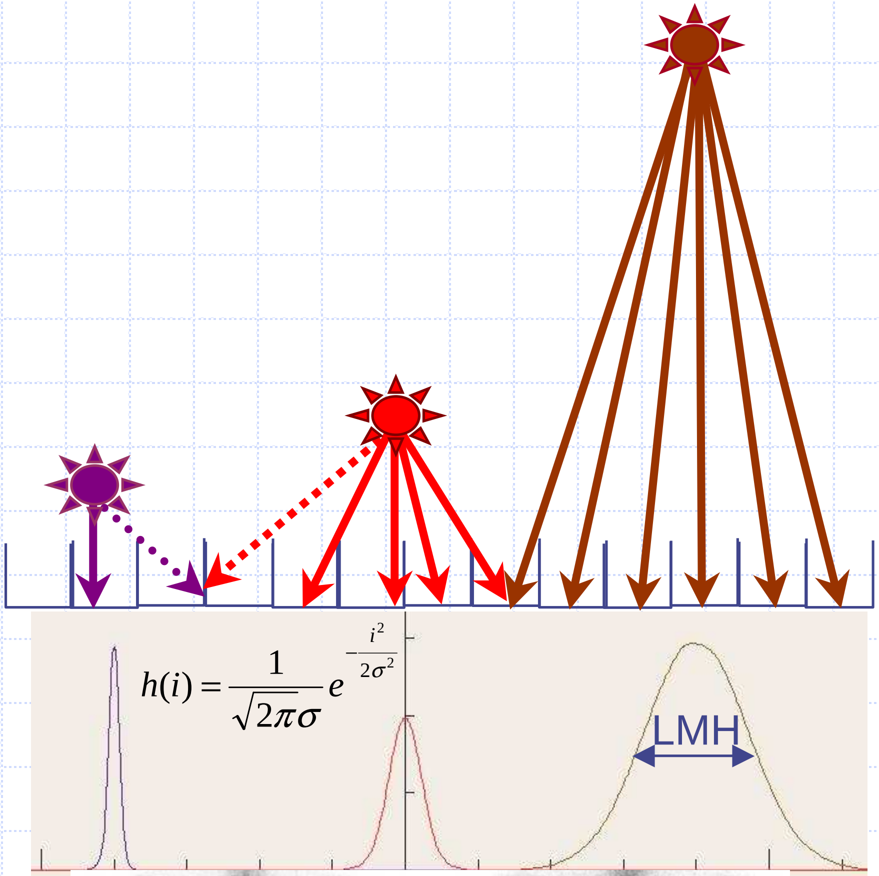
$h_i \star h_c \approx$ Gaussienne

Quantitative Analysis in Nuclear Medicine Imaging, Springer US. 2006; CE Metz. Phys Med Biol 1990; 35.

Linéarité entre LMH et D

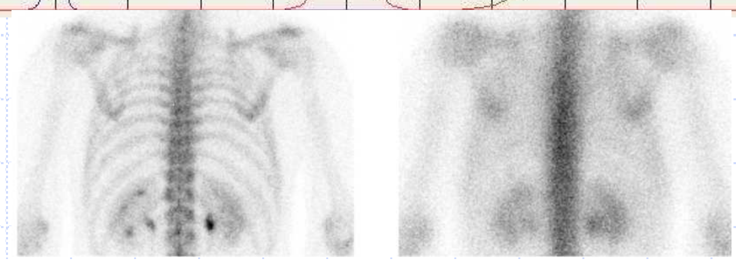
$h_i \star h_c \approx$ Gaussienne





$$LMH = 2.\sqrt{2.\ln 2}\sigma = kD + k'$$

Utilité d'une déconvolution (non stationnaire)

contact  50 cm

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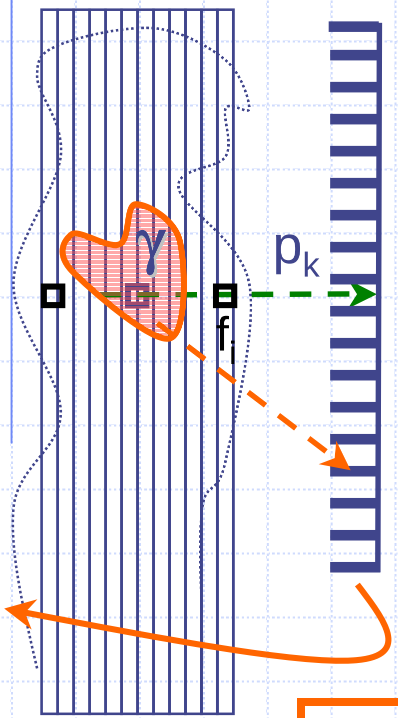
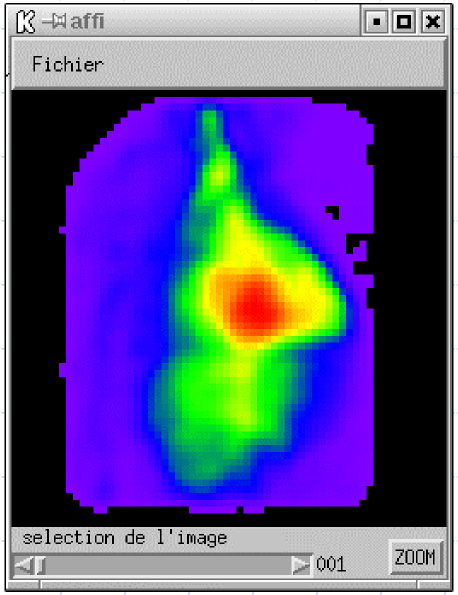
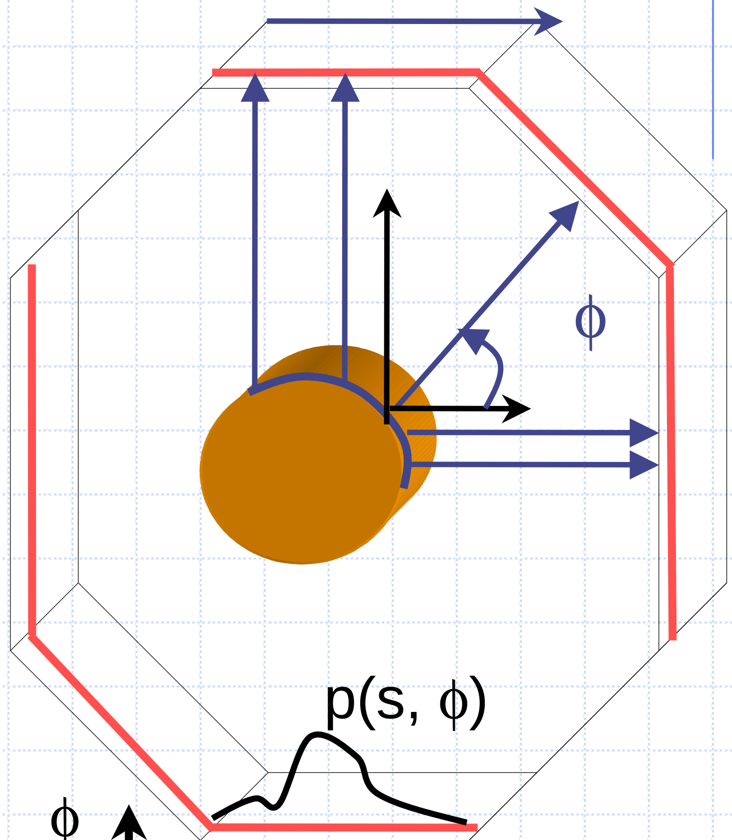
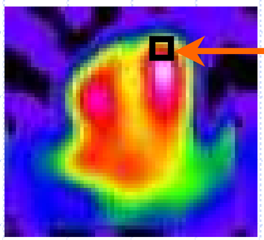


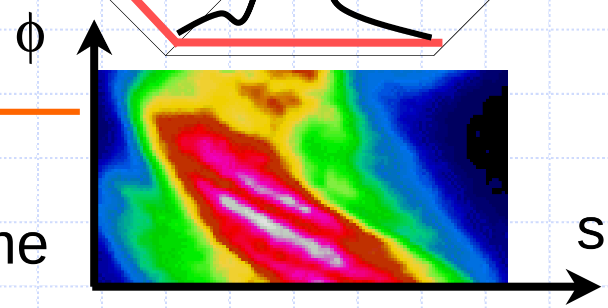




$$p_k = \sum_i r_{k,i}.f_i \Leftrightarrow \vec{p} = R.\vec{f}$$

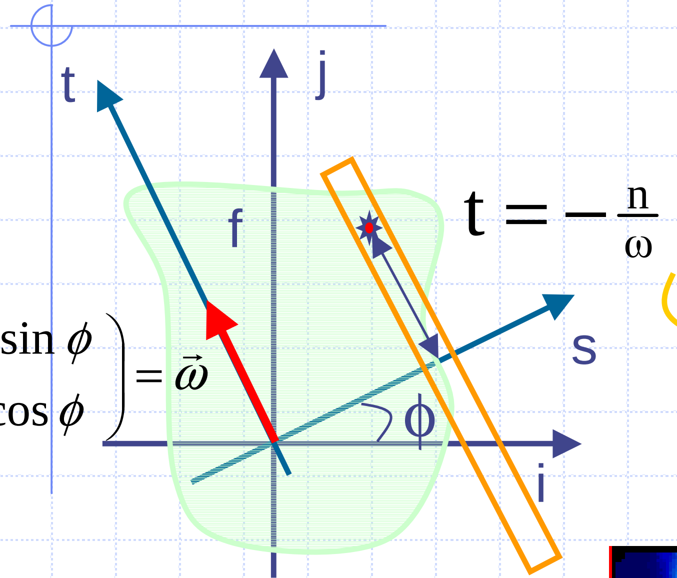


$f_i = ?$



sinogramme

Déconvolution (hors synchro ECG)



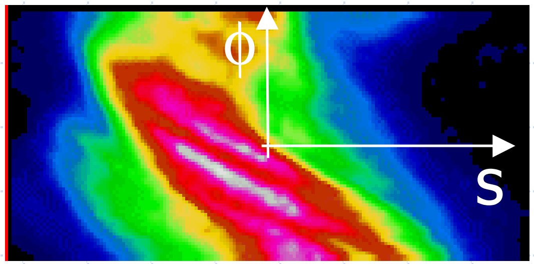
$$\begin{pmatrix} -\sin\phi \\ \cos\phi \end{pmatrix} = \vec{\omega}$$

$$t = -\frac{n}{\omega}$$

sources à t mm du centre de rotation

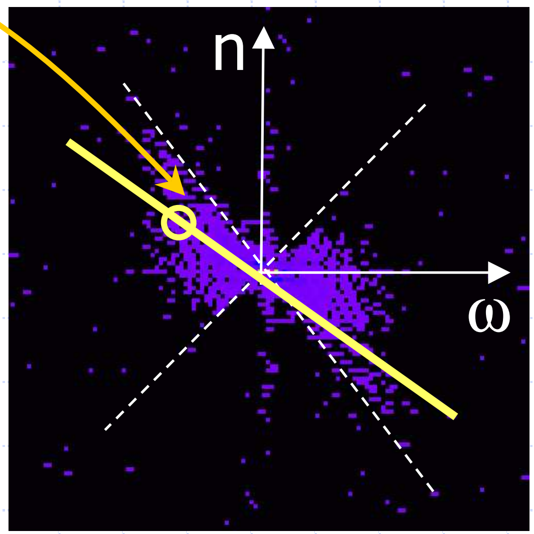
⇩

signal ≈ sur la droite $n=-\omega.t$ de la TF2 du sinogramme



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

$\xrightarrow{TF_2}$

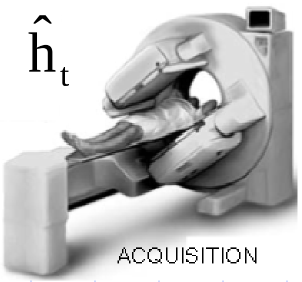


$$\hat{\hat{p}}_c(\sigma, n)$$

P.R Edholm, RM Lewitt, B Lindholm. « Novel Properties of the Fourier decomposition of the sinogram». Proc. SPIE, 671,8-18 (1986)

Principe fréquence-distance

$\hat{p}$ → $\hat{h}_t$

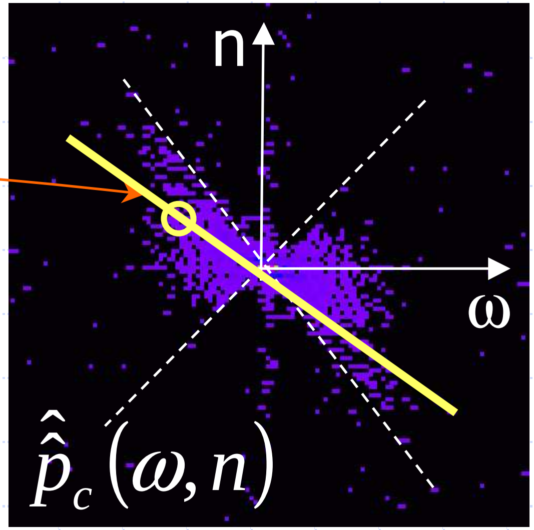


→ $\hat{p}_c = \hat{h}_t \cdot \hat{p}$ → $\dfrac{1}{\hat{h}_t}$ → $\hat{p}$

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

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

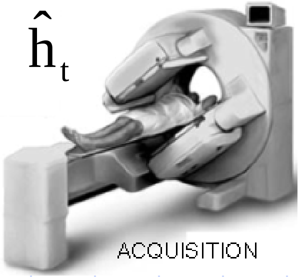
$$\hat{p}(\omega, n) = \frac{1}{\hat{h}_{-\frac{n}{\omega}}(\omega)}.\hat{p}_c(\omega, n)$$



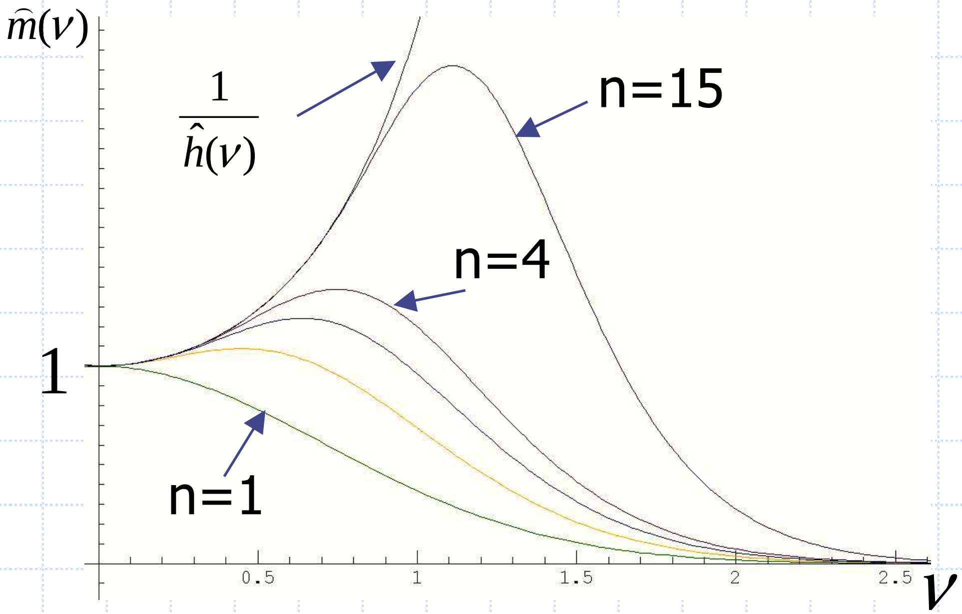
W Xia et al, IEEE TMI 1995;14-Hawkins et al. IEEE TMI 1988;7-Lewitt et al. Proc SPIE 1989; 1092-Glick et al. IEEE TMI 1994;13.

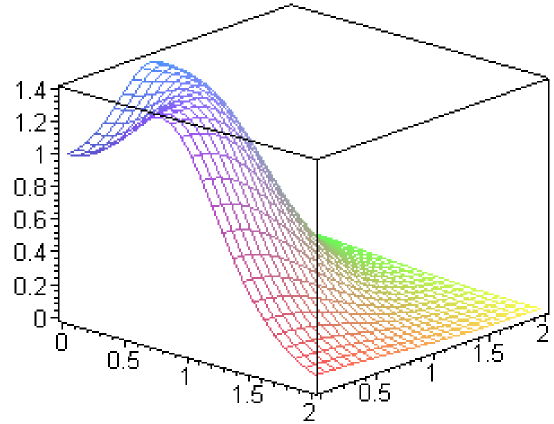
Filtre de déconvolution de Metz

$\hat{p}$ → $\hat{h}_t$



→ $\hat{\hat{p}}_c = \hat{h}_t \cdot \hat{p}$ → $\dfrac{1}{\hat{h}_t}$ → $\hat{\hat{p}}$



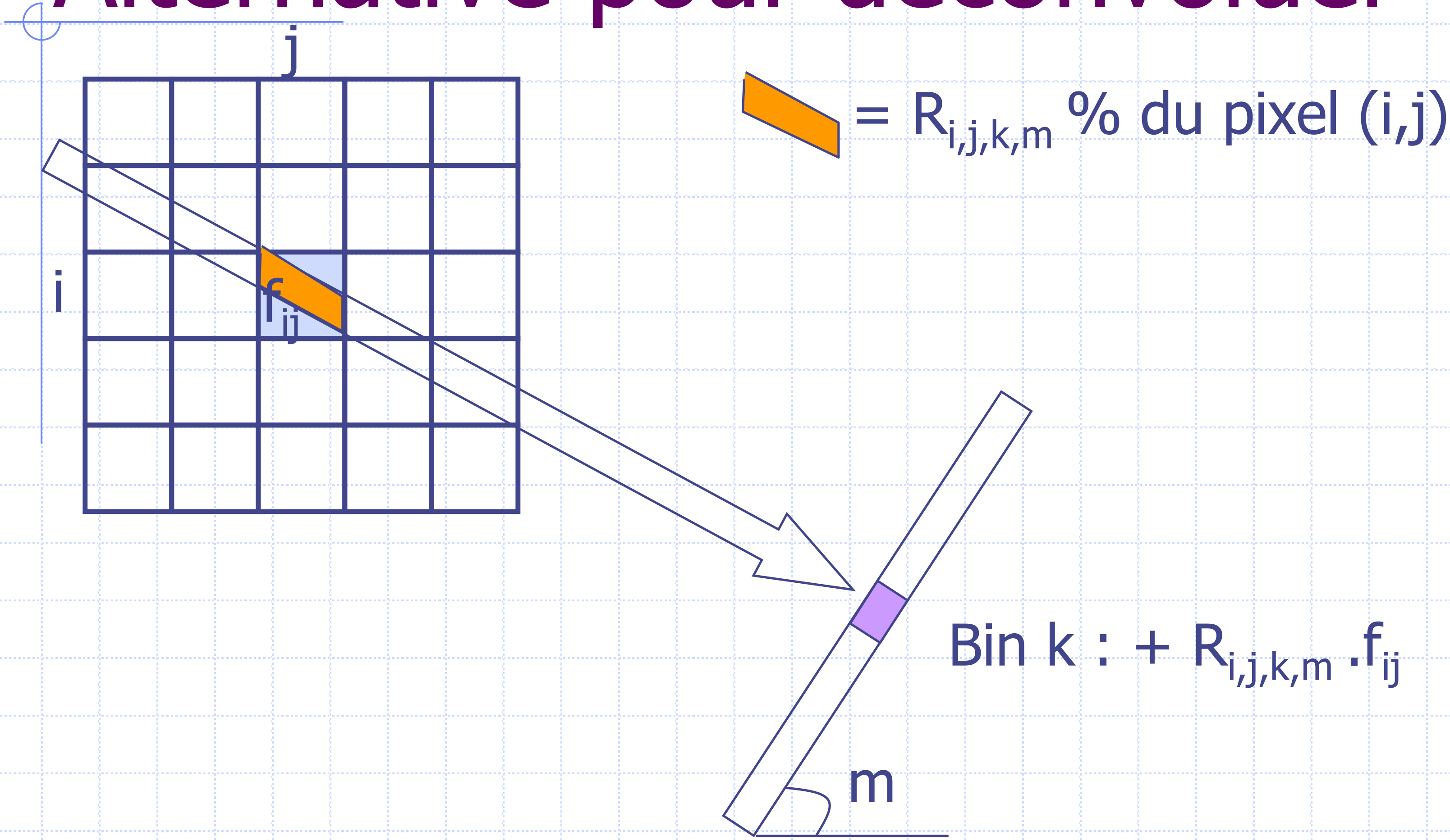


$$\hat{m}(\nu,\nu') = \frac{1-\left[1-\hat{h}(\nu,\nu')^2\right]^n}{\hat{h}(\nu,\nu')}$$

$$n = 0{,}834.\ln(C) - 7.774$$

King et al. JNM 1983;24 et Metz et al. JNM 73;15

Alternative pour déconvoluer



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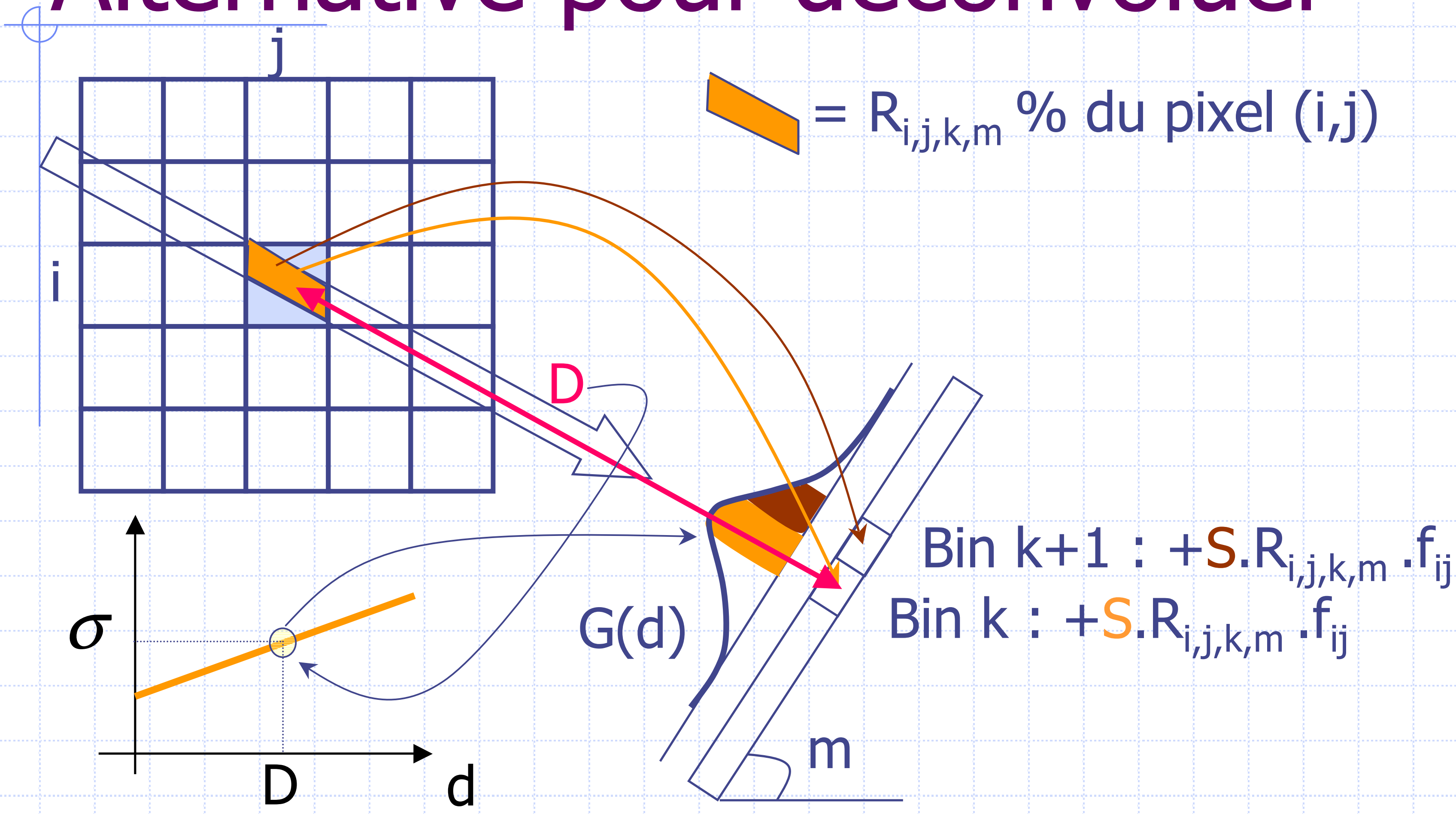
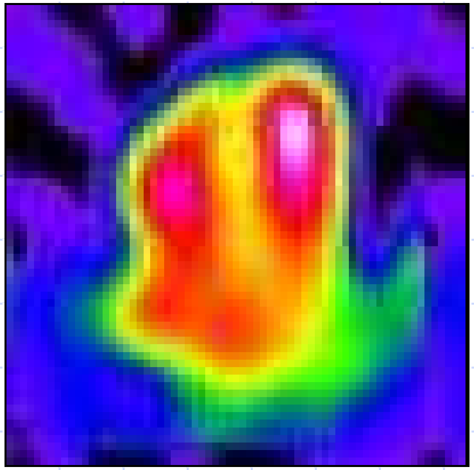


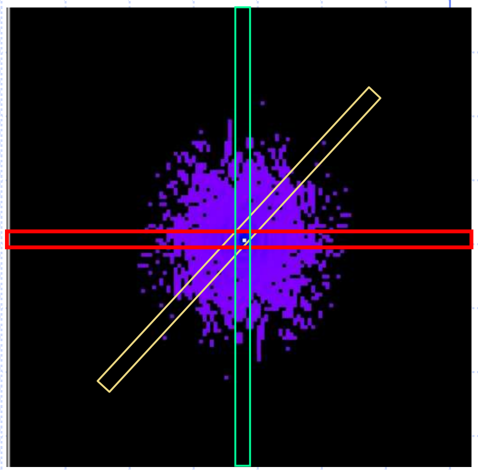
Tsui et Floyd, IEEE Trans Nucl Sci 1988;35 – A Formiconi, Phys Med Biol 1989;34 – Penney, IEEE TNS 1990;37 - McCarthy, IEEE TMI 1991;10 – Zeng, IEEE TNS 1991;38-1992;39 – Bechman, IEEE TNS 1993; 40

Solvabilité du problème tomographique

Aspect analytique: R bijectif d'inverse continu



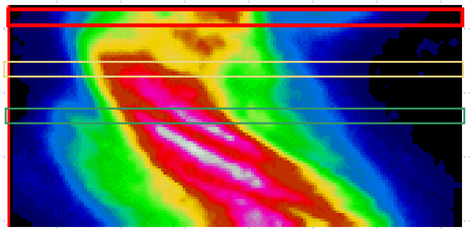
$$\hat{p}_{\vec{\omega}^{\perp}}(\sigma) = \hat{f}\left(\sigma.\vec{\omega}^{\perp}\right)$$

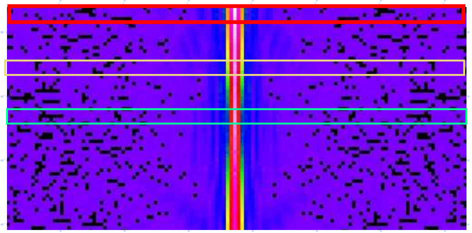


f ⟺ TF ⟺ $\hat{f}$

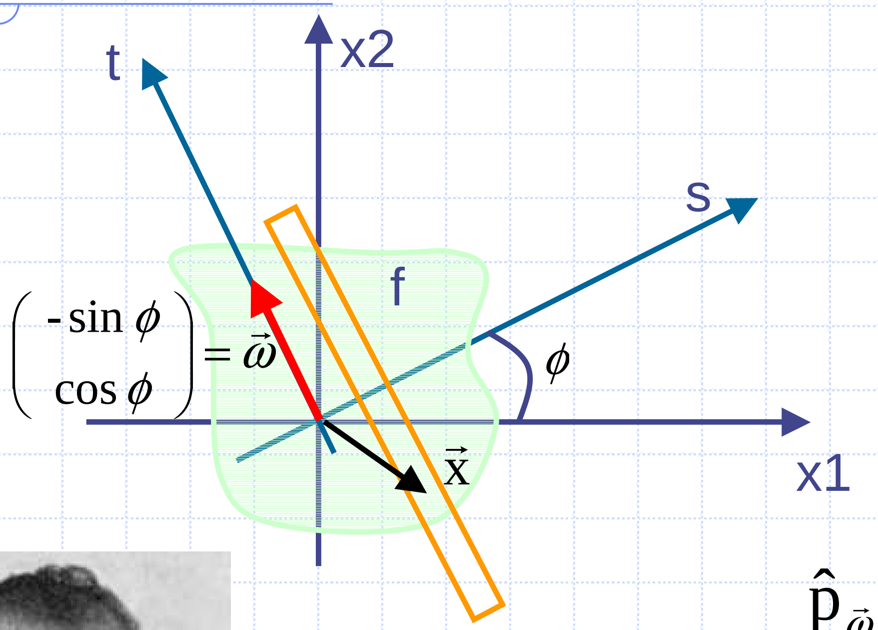
R ⇕ ‖ ⇕

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





Démonstration



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

J. Radon
1887-1956

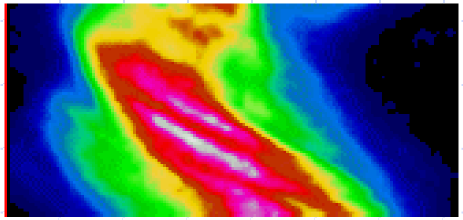
$$\hat{p}_{\vec{\omega}}(\sigma) = \int_s \int_t f(s\vec{\omega}^{\perp} + t\vec{\omega})\, e^{-i.s.\sigma}\, dt\, ds$$

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

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

Johann Radon, ¨ Uber die Bestimmung von Funktionen durch ihre Integralwerte langs gewisser Mannigfaltigkeiten, Ber. Verh. Sach. Akad. 69 (1917), 262–77.

Rétroprojection filtrée



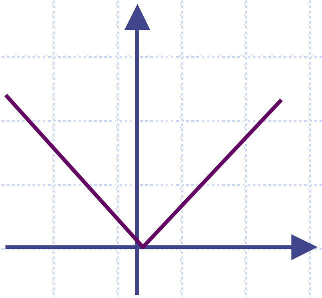
$$f(\vec{x}) = (R^{*} p^{f})(\vec{x})$$

Projections sur 180°

$p_{\vec{\omega}}$

↓

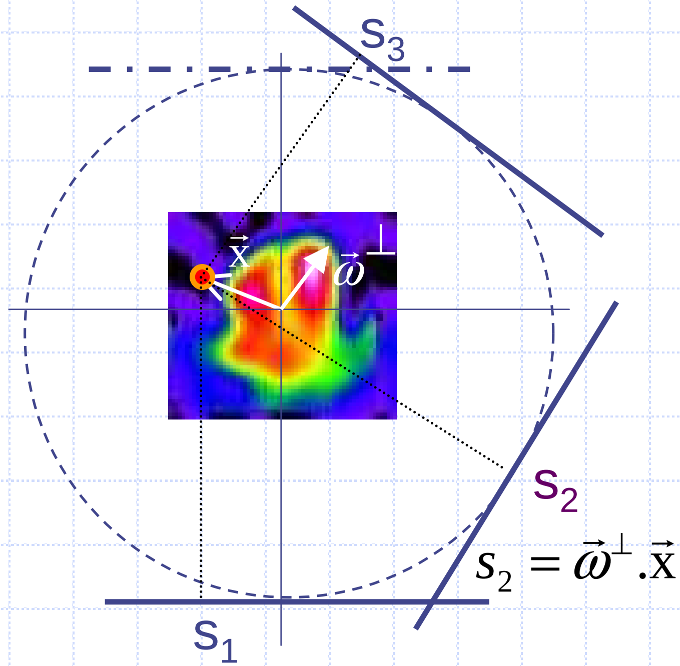
$\hat{p}_{\vec{\omega}}$ abs



x

↓

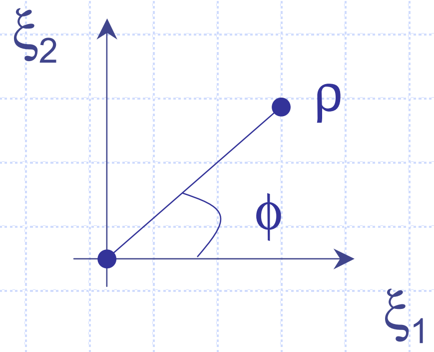
$$TF_s^{-1}\left[\hat{p}_{\vec{\omega}} \,.\, abs\right] = p^{f}_{\vec{\omega}}$$



Démonstration

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

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



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

$$\| $$

$$\underbrace{TF_s^{-1}\left[\hat{p}_{\vec{\omega}^{\perp}}\,.\,abs\right]}_{p^{f}_{\vec{\omega}}}\left(\vec{\omega}^{\perp}.\vec{x}\right)$$

J. Radon

1887-1956

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

Solvabilité du problème tomographique

Aspect discret : le problème est moins simple !

- ✓ R surjectif ? : $\,^t R.R\vec{f} = \,^t R.\vec{p} \Leftarrow \bar{f} = \arg\min_f \left\| \vec{p} - R\vec{f} \right\|^2$
- ✓ R injectif ? :
- ✓ Norme de R^{-1} ?

  R^{-1} continue mais $||R^{-1}||$ grande

$\Leftarrow$ choix parmi les solutions possibles

$$\frac{\left\|\delta\vec{f}\right\|}{\left\|\vec{f}\right\|} \leq \kappa(R)\frac{\left\|\delta\vec{p}\right\|}{\left\|\vec{p}\right\|} \quad \text{où} \quad \kappa(R) = \left\|R\right\| \left\|R^{-1}\right\| = \frac{\lambda_{max}}{\lambda_{min}} >> 1$$

Gradient Conjugué

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

Initialisation :

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

$$\omega^{j} = \frac{\left\|\vec{r}^{\,j}\right\|^2}{\left\langle \vec{d}^{\,j} \middle| R^{*}.R.\vec{d}^{\,j} \right\rangle}$$

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

$$\vec{d}^{\,j+1} = \vec{r}^{\,j+1} + \frac{\left\|\vec{r}^{\,j+1}\right\|^2}{\left\|\vec{r}^{\,j}\right\|^2}.\vec{d}^{\,j}$$

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

Gilbert P Iterative methods for the three-dimensional reconstruction of an object from projections. J. Theor. Biol. 1972; 36:105-17

Estimation de κ(R) par Rf=p

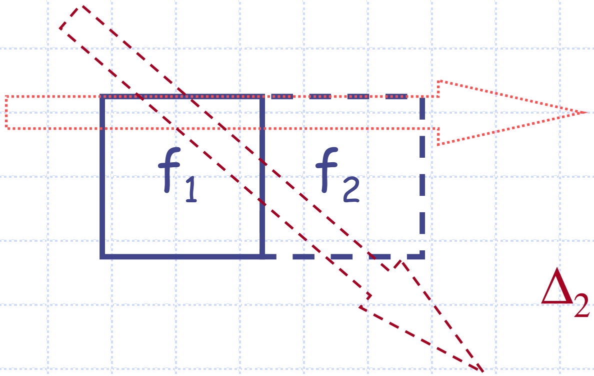
- Gradient conjugué $\omega^j = \dfrac{\left\| \vec{r}^{\,j} \right\|^2}{\left\langle \vec{d}^{\,j} \middle| R^*.R.\vec{d}^{\,j} \right\rangle}$, $\beta^j = \dfrac{\left\| \vec{r}^{\,j+1} \right\|^2}{\left\| \vec{r}^{\,j} \right\|^2}$
- Approximation de Galerkin

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

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

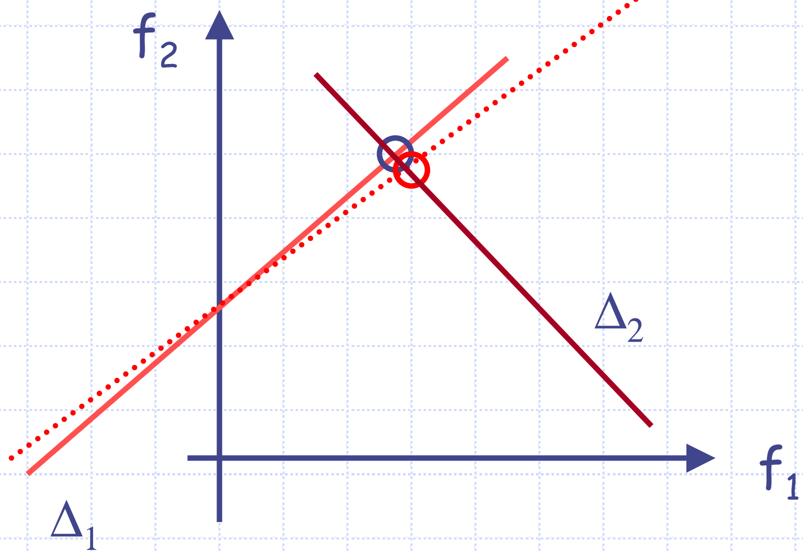
www.inma.ucl.ac.be/~vdooren/Krylov.pdf
Analyse numérique matricielle appliquée à l'art de l'ingénieur. P. Lascaux & R. Théodor (II), p. 516, 1987. MASSON
D. Mariano-Goulart, P. Maréchal, S. Gratton, et al. Comput. Med. Imaging & Graphics 2007; 31 : 502-509

Approche intuitive (I)

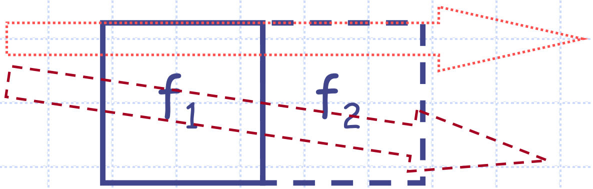


$\Delta_1 : p_1 = r_{1,1}\, f_1 + r_{1,2}\, f_2$

$\Delta_2 : p_2 = r_{2,1}\, f_1 + r_{2,2}\, f_2$

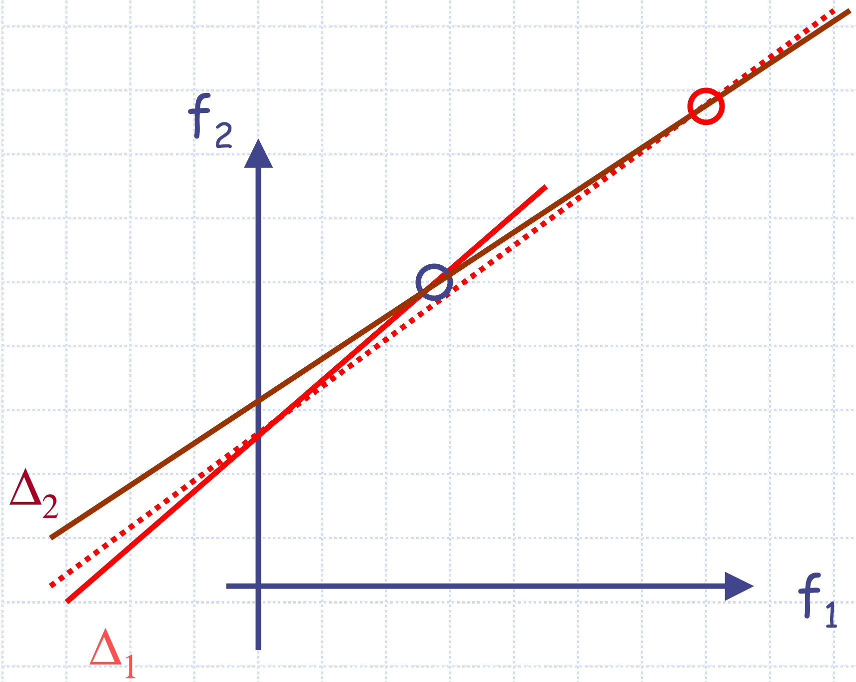


Approche intuitive (II)



$\Delta_1 : p_1 = r_{1,1} f_1 + r_{1,2} f_2$

$\Delta_2 : p_2 = r_{2,1} f_1 + r_{2,2} f_2$



$64^2 = 4\,096$

$128^2 = 16\,384$

$256^2 = 65\,536$

$512^2 = 262\,144$

Exemple

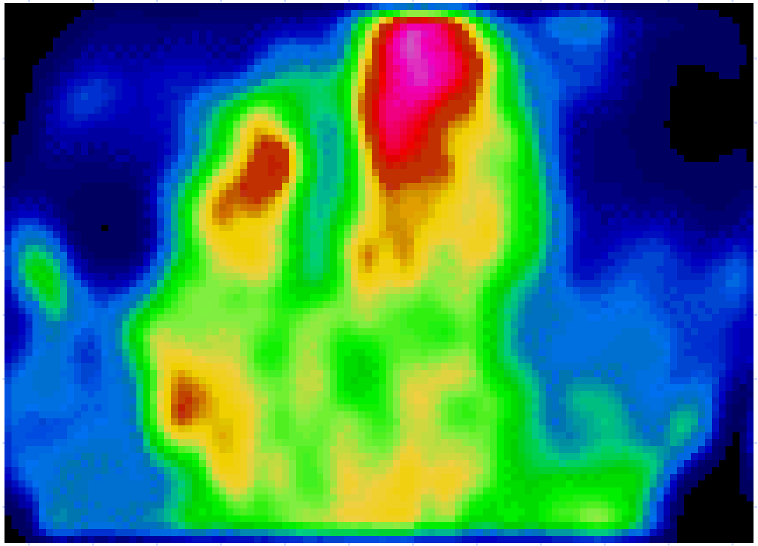
Matrice de Wilson

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

$$\mathrm{Sp(R)} \approx \{0{,}010;\ 0{,}843;\ 3{,}858;\ 30{,}289\} \Rightarrow \kappa(\mathrm{R}) = \frac{30{,}289}{0.010} = 2984$$

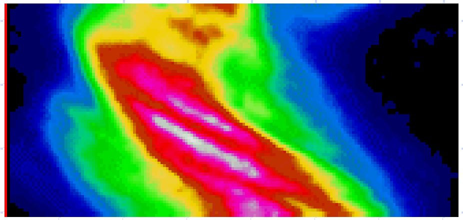
$$\begin{pmatrix} 10 & 7 & 8 & 7 \\ 7 & 5 & 6 & 5 \\ 8 & 6 & 10 & 9 \\ 7 & 5 & 9 & 10 \end{pmatrix} \begin{pmatrix} 9{,}2 \\ -12{,}6 \\ 4{,}5 \\ -1{,}1 \end{pmatrix} = \begin{pmatrix} 32{,}1 \\ 22{,}9 \\ 33{,}1 \\ 30{,}9 \end{pmatrix}$$

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



100 itérations de MLEM

Régularisation de la RPF



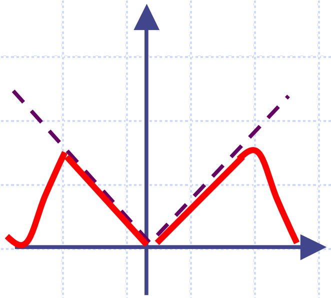
$$f(\vec{x}) = (R^* p^f)(\vec{x})$$

Projections sur 180°

$p_{\vec{\omega}}$

↓

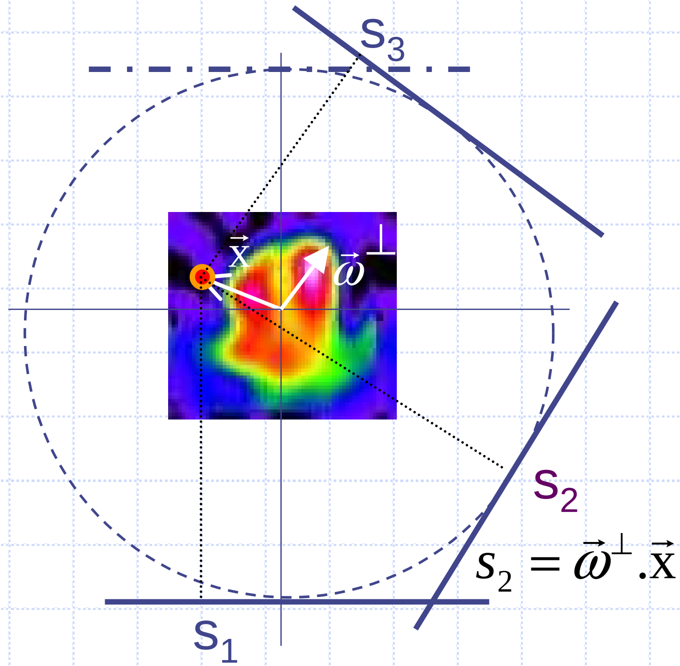
$\hat{p}_{\vec{\omega}}$ abs



x

↓

$$TF_s^{-1}\left[\hat{p}_{\vec{\omega}} . \mathrm{abs}\right] = p^f_{\vec{\omega}}$$



Régularisation algébrique

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

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

Adéquation aux données
Surjectivité du problème inverse

Régularisation
injectivité

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

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

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

Régularisation MAP-EM-OSL

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

$$\tilde{\vec{f}} = \arg\min_{\vec{f}} \left[-\log P(\vec{p}/\vec{f}) - \log P(\vec{f}) \right]$$

Adéquation aux données

régularisation

$$\log \prod_i \frac{e^{-\tilde{p}_i}\, \tilde{p}_i^{\,p_i}}{p_i!}$$

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

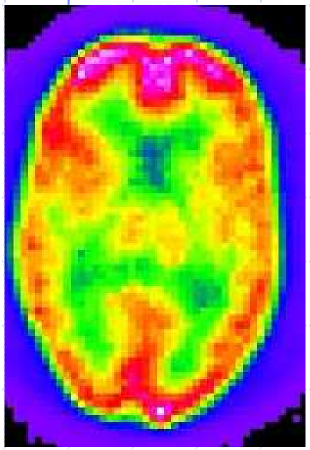
$$f_i^{n+1} = f_i^{n}.\frac{1}{\sum_{l'=1}^{P} r_{l',i}} \sum_{l=1}^{P} r_{l,i} \frac{p_l}{\sum_{s=1}^{N} r_{l,s} f_s^{n}}$$

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

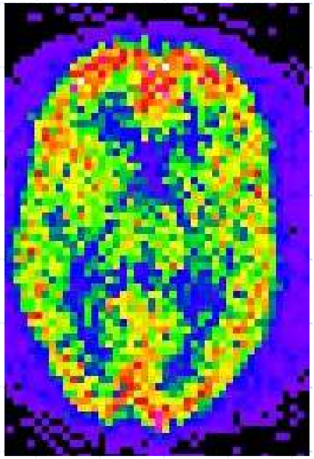
$$\partial U = \sum_{f_k \in V(f_i)} w_{i,k}.\frac{\partial V}{\partial r}(f_i - f_k)$$

Dempster A et al. Maximum likelihood from incomplete data via the EM algorithm. J R Stat Soc 1977;39:1-38.
Hudson H et al. .Accelerated image reconstruction using ordered subsets of projection data. IEEE TMI 1994;13:601-9.
Green PJ. Bayesian reconstructions from emission tomography data using a modified EM algorithm. IEEE Trans Med Imaging 1990;9:84-93

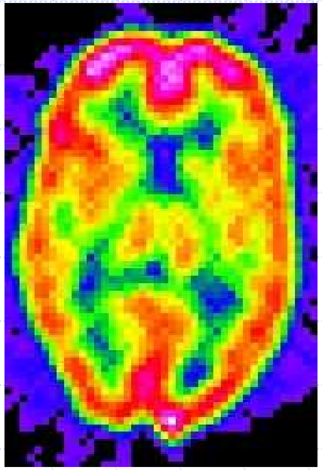
Comparaison MLEM-GC-FRECT



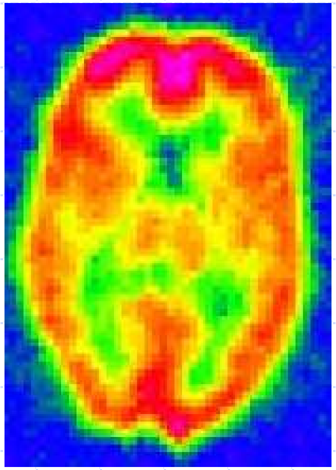
MLEM 6



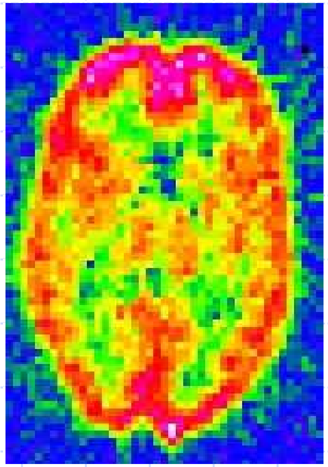
MLEM 200



FRECT 34 (CV)



GC 6



GC 16

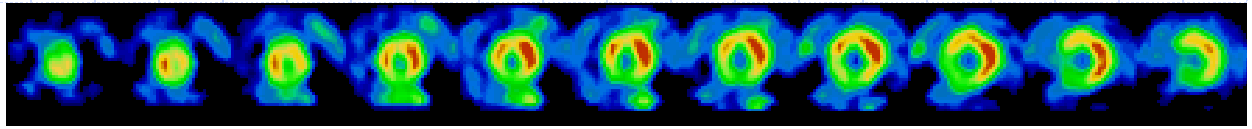
$$E(f) = \left\| TF_s^{-1}(B.\hat{p}) - Rf \right\|^2 + \left\| (1 - B).\hat{f} \right\|^2$$

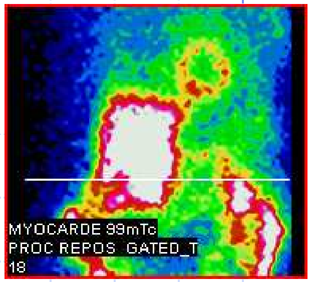
D Mariano-Goulart, P Maréchal, S Gratton, L Giraud, M Fourcade. *Comput Med Imaging Graph* 07 & *Lect Notes Comput Sci*. 04

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 et Analyse

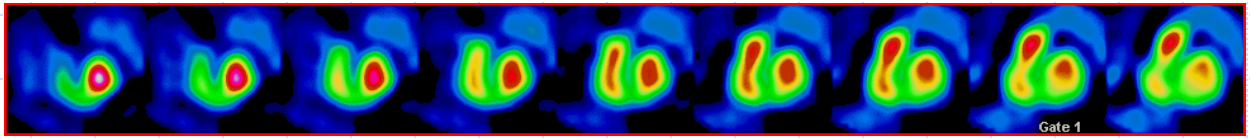
- Scintigraphie myocardique (MIBI ou MIBG)

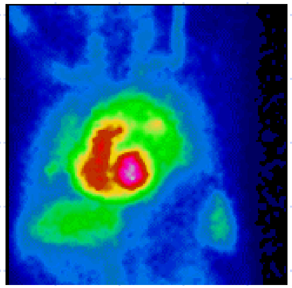




  - Faible rapport signal/bruit
  - Activités extra cardiaques
  - Ventricule gauche ≈ ellipsoïde tronqué

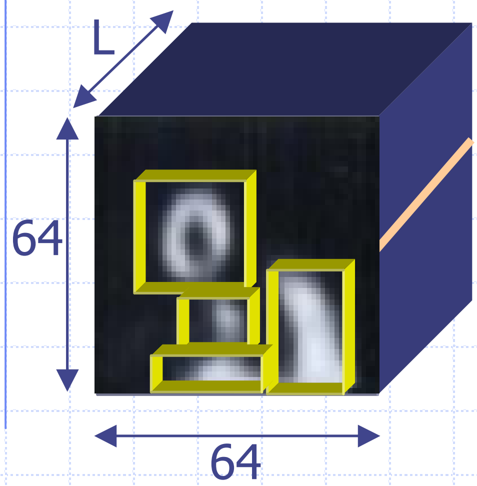
- Tomoventriculographie isotopique





  - 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

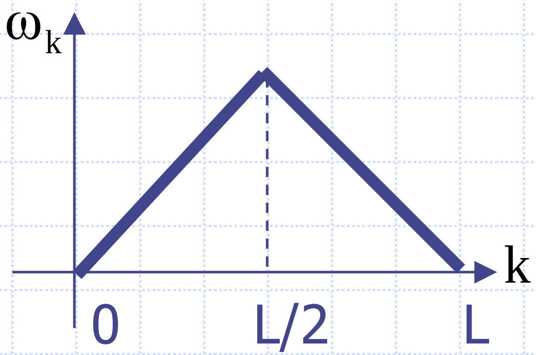
$$S(i,j) = \sum_{k=1}^{L} \omega_k . SA(i,j,k)$$

② ? Maxima locaux de S(i,j)

- supérieur seuil et au moins 4-connexes

③ Transformée de Hough

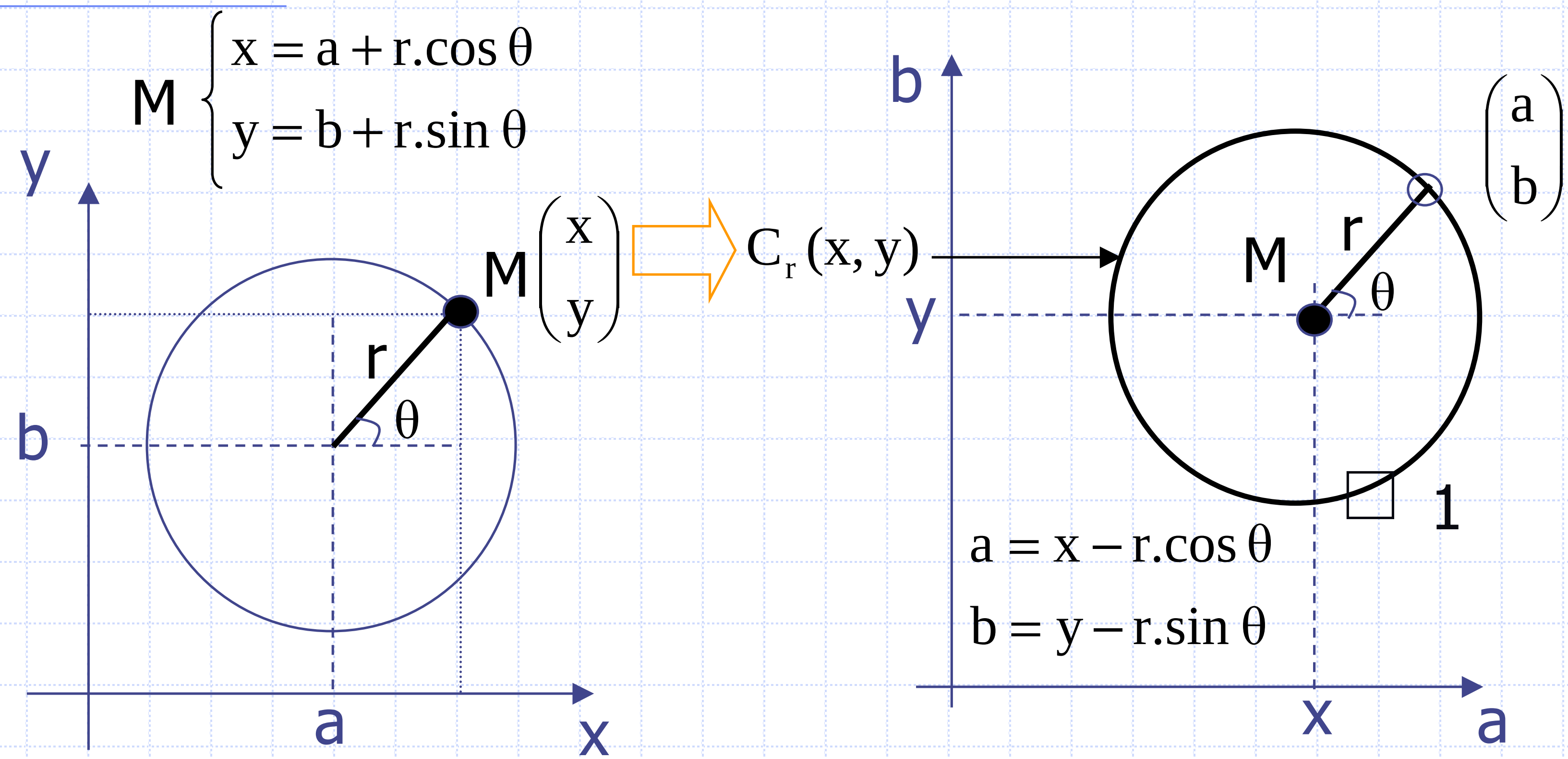
- ? cercles dans un nuage de points



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

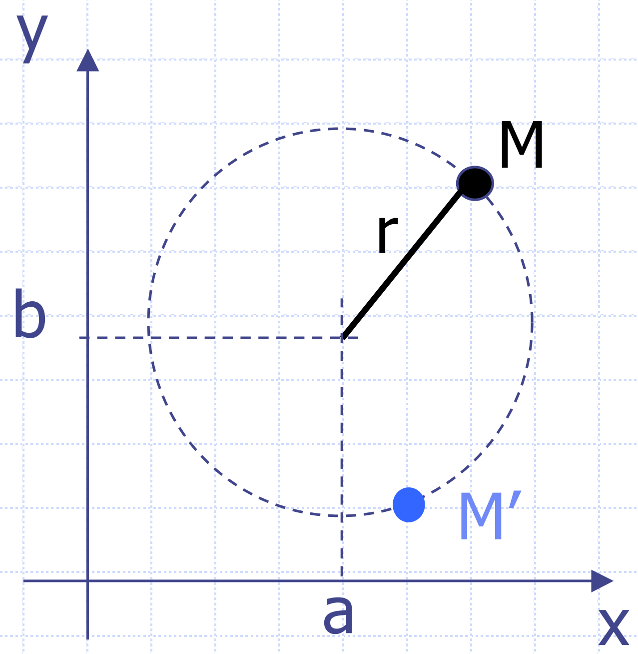
QPS® : TRANSFORMATION DE HOUGH

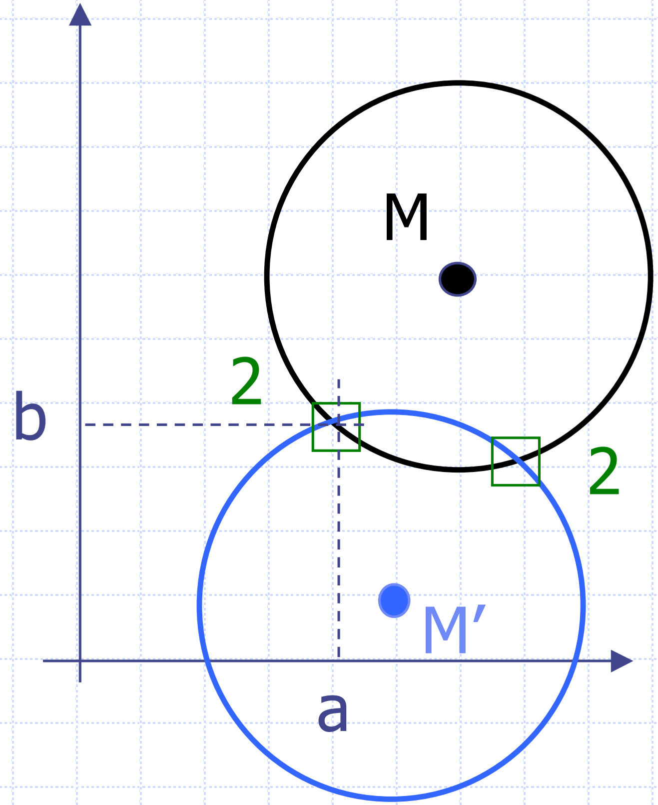


à tout **point** M du cercle(O,r) on associe un **cercle** (M,r)

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





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http://homepages.inf.ed.ac.uk/rbf/BOOKS/BANDB/bandb.htm

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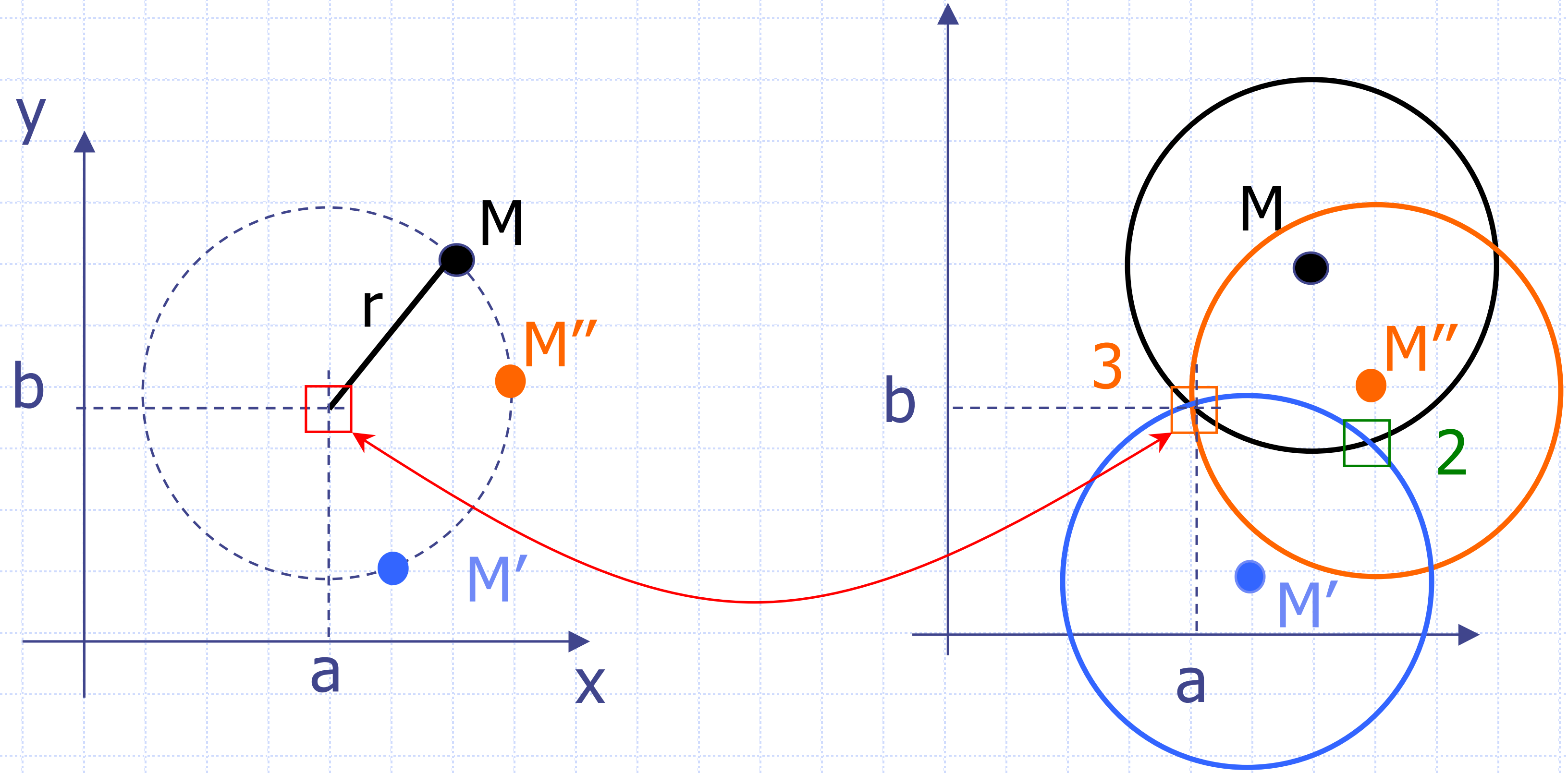

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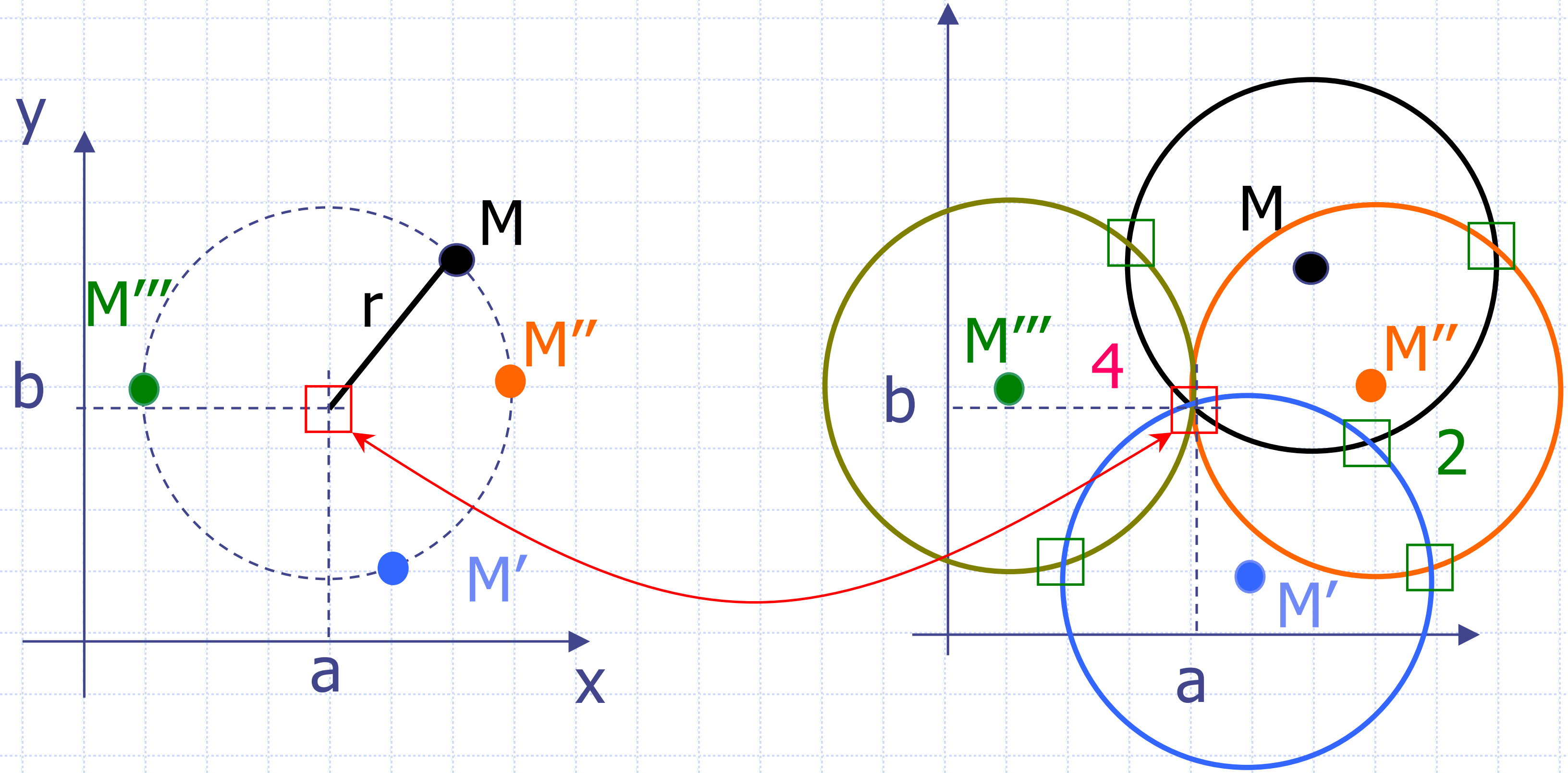
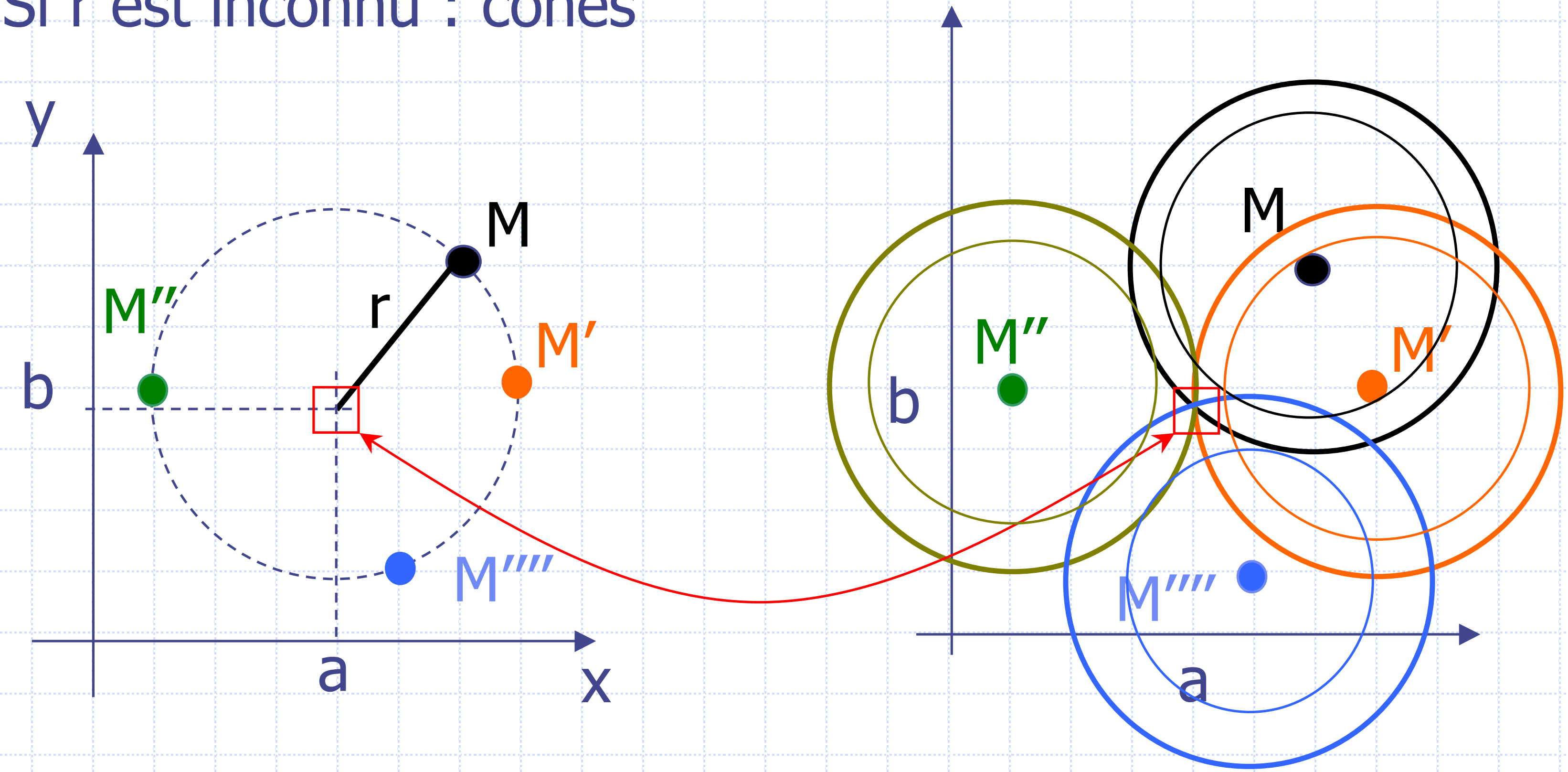


D. Ballard, C. Brown. Computer vision. Englewood Cliffs, NJ: Prentice-Hall; 1982.
http://homepages.inf.ed.ac.uk/rbf/BOOKS/BANDB/bandb.htm

QPS® : TRANSFORMATION DE HOUGH

Si r est inconnu : cônes

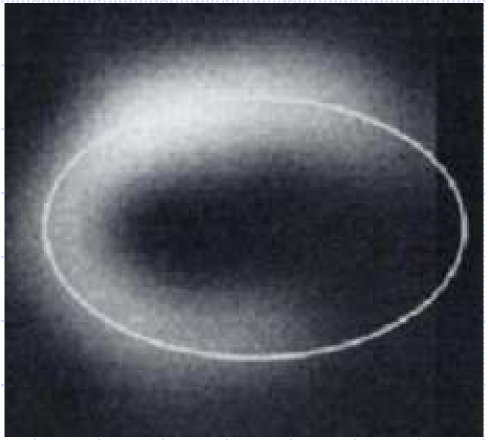


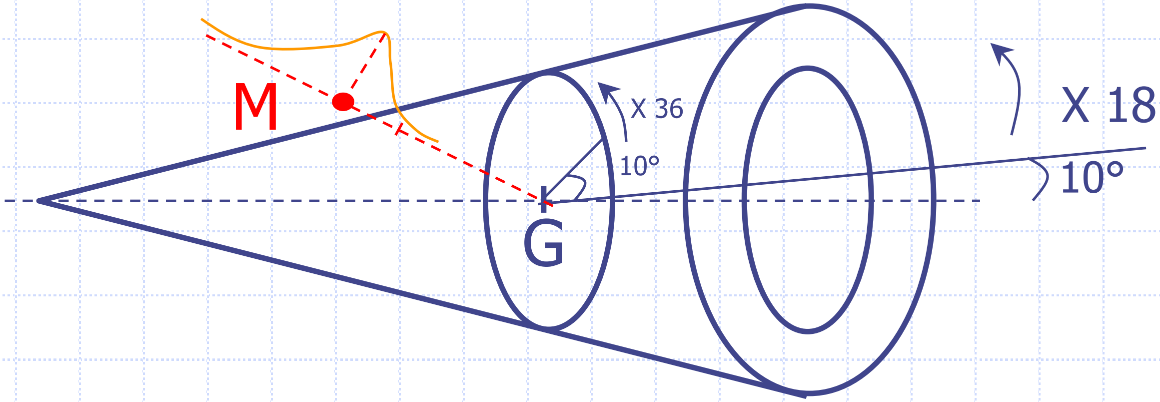
? Extrema dans l'espace des paramètres 3D: (a,b,r)

Puis QPS choisit le cercle à plus fort comptage moyen sur la circonférence par rapport au comptage au centre

QPS® : MYOCARDE

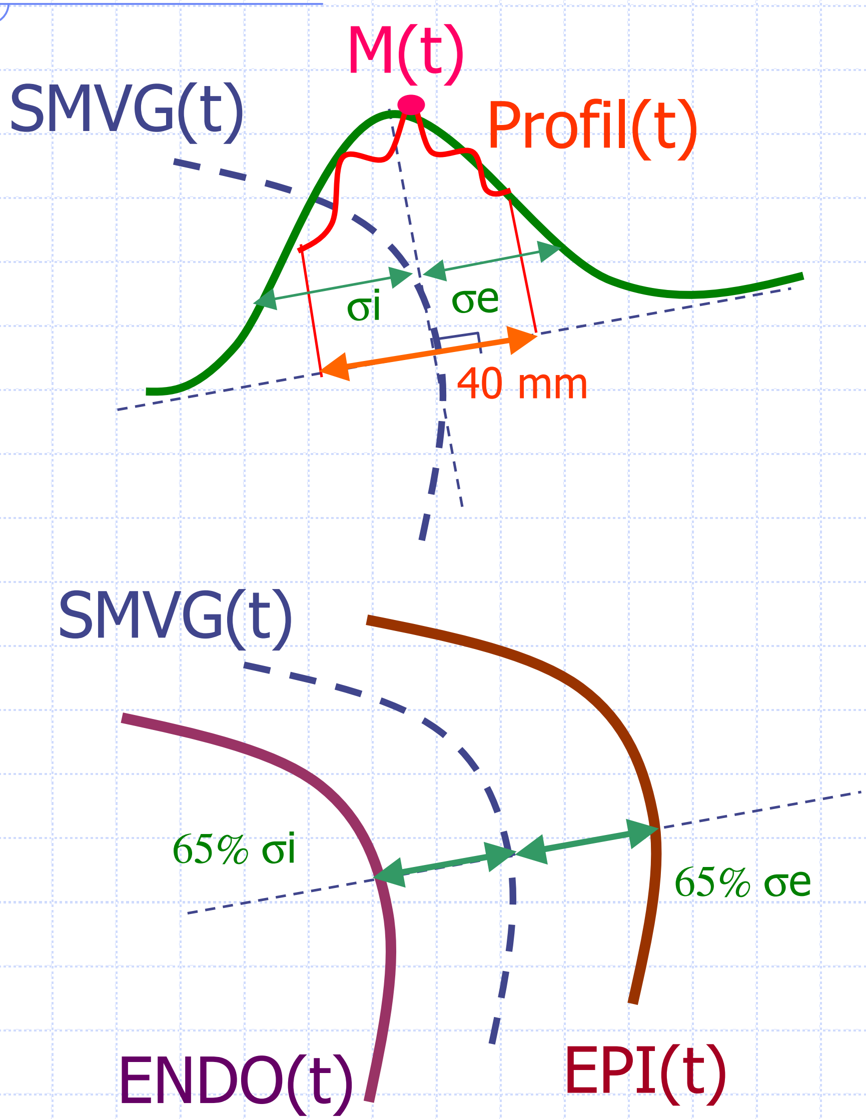
① G : Centre de Gravité du masque binaire
② Données = Volume SPECT x Masque
③ Extraction de 18x36 profils d'activité
④ Surface médio VG = maxima des profils M
⑤ Fit ellipse 3D→ axe VG → G ← proj.(G)/axe

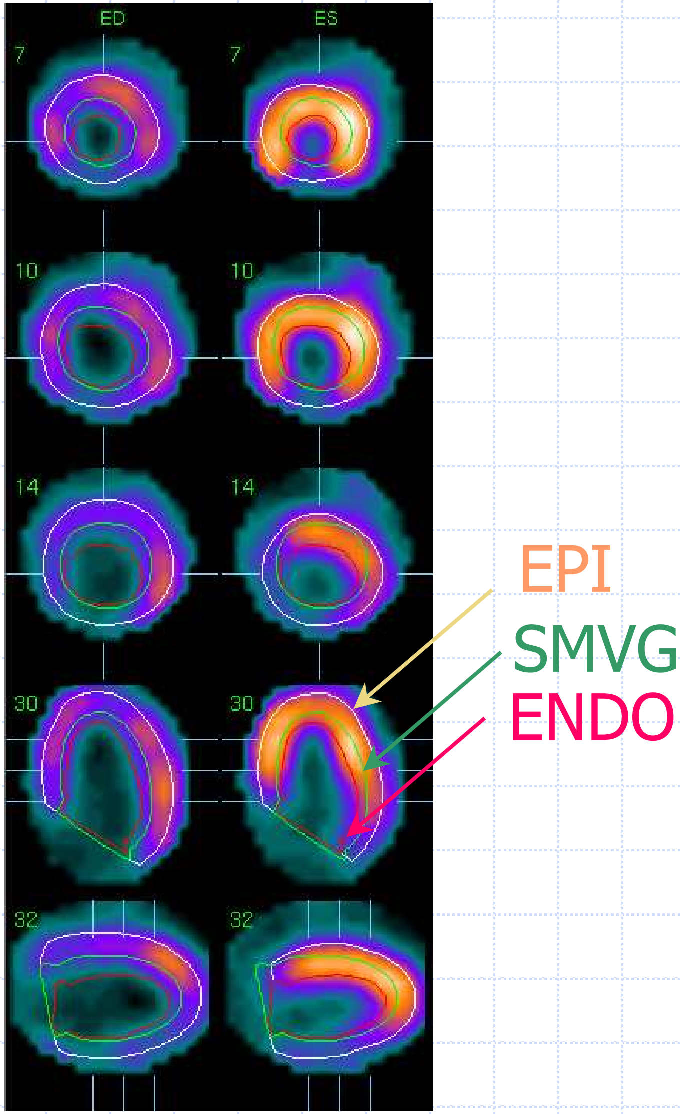




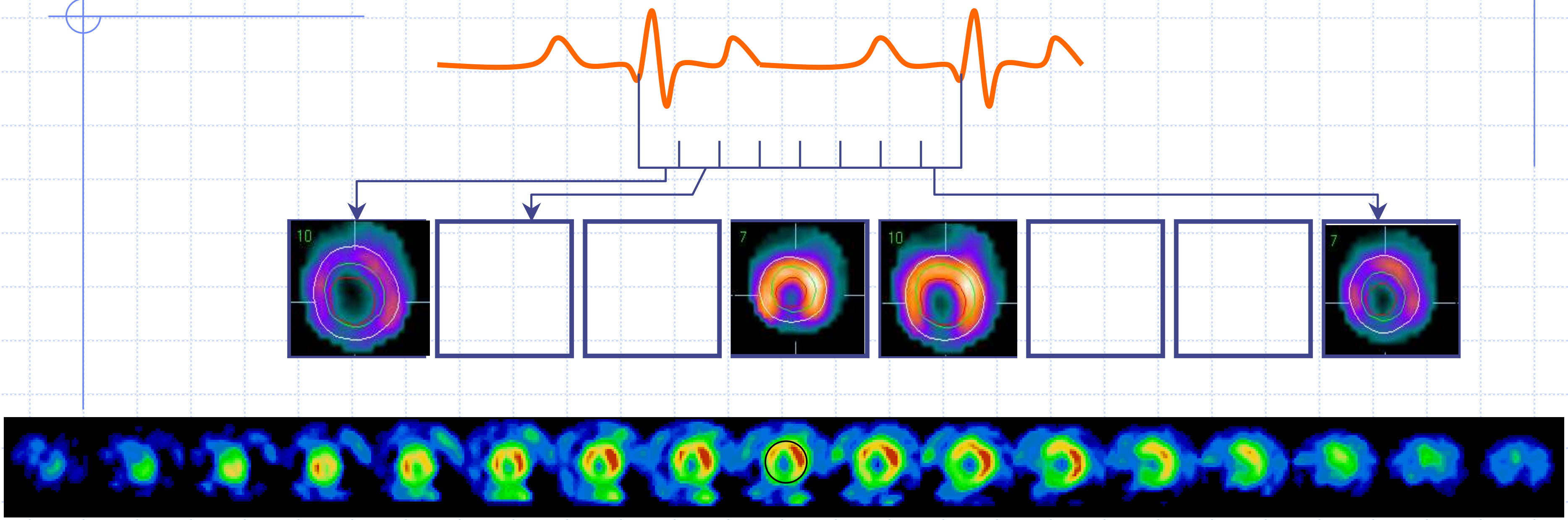
JC Cauvin et al. Eur J Nucl med 1992;19:1032-1037
G. Germano et al. JNM 1995;36:1107-14 ; 1995;36:2138-47 et JNM 2000;41:712-19

QPS® : SEGMENTATION DU MYOCARDE



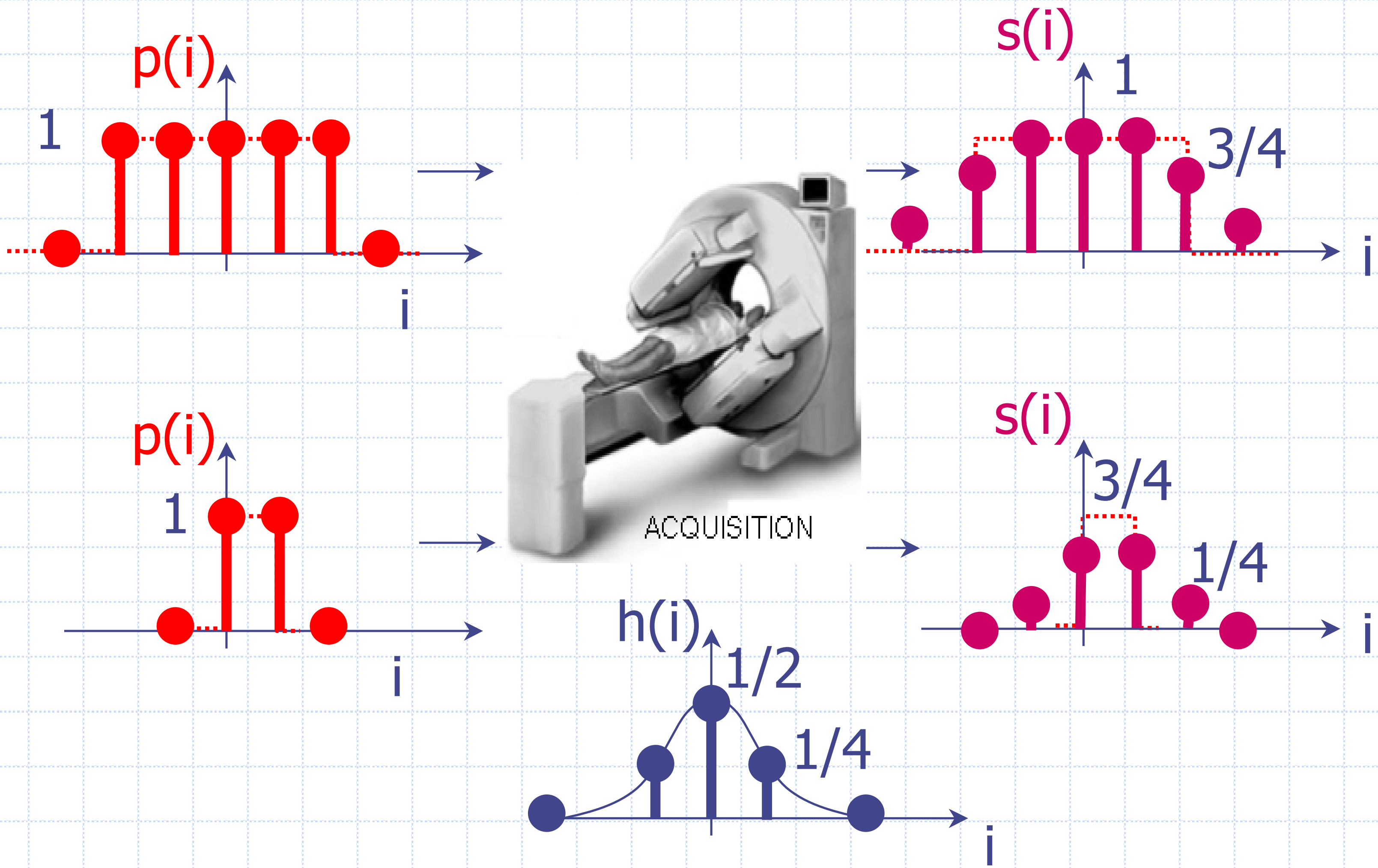


MOUVEMENT MYOCARDIQUE ?

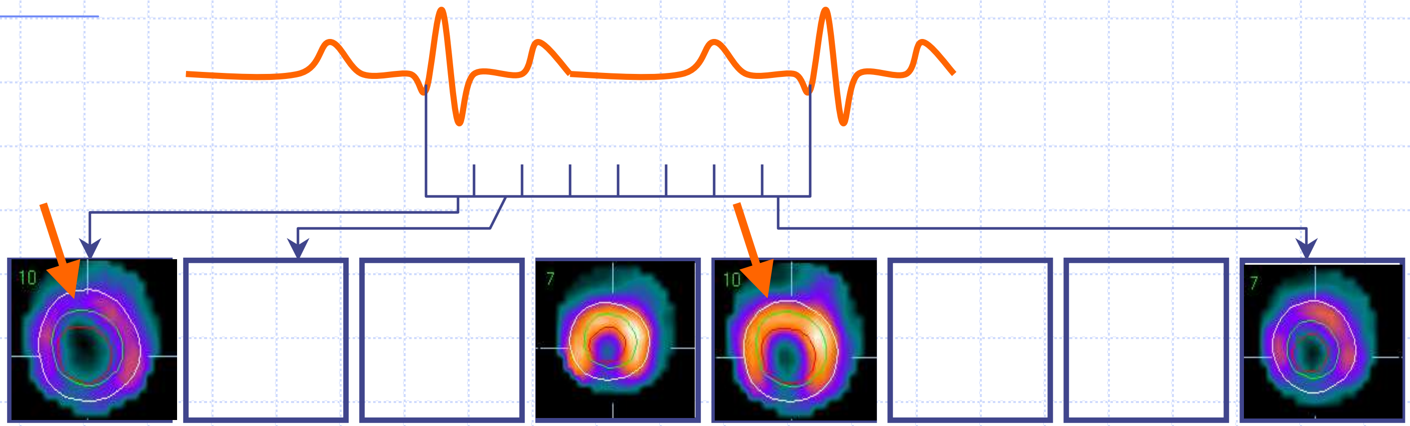


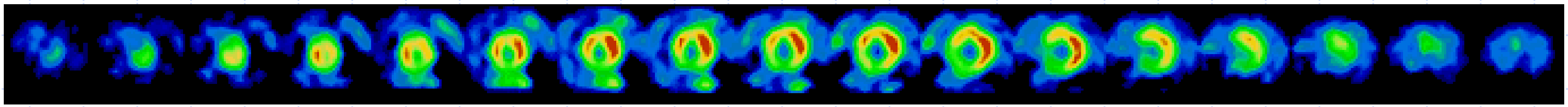
mais LMH ≈ 15 mm ≈ mouvement
qui est donc peu ou pas visible.

Rappel : Effet de volume partiel



EPAISSISSEMENT SYSTOLIQUE



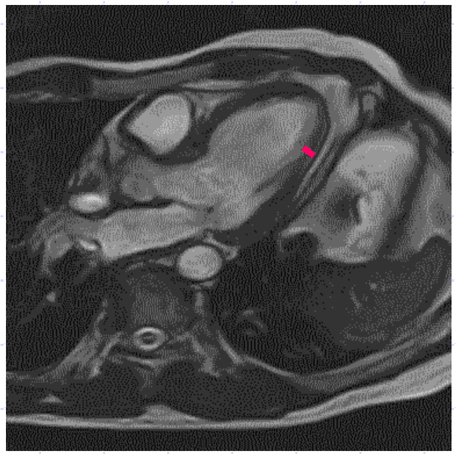


Epaisseur systolique ≈ 30 mm

↳ 100 % de l'activité mesurée

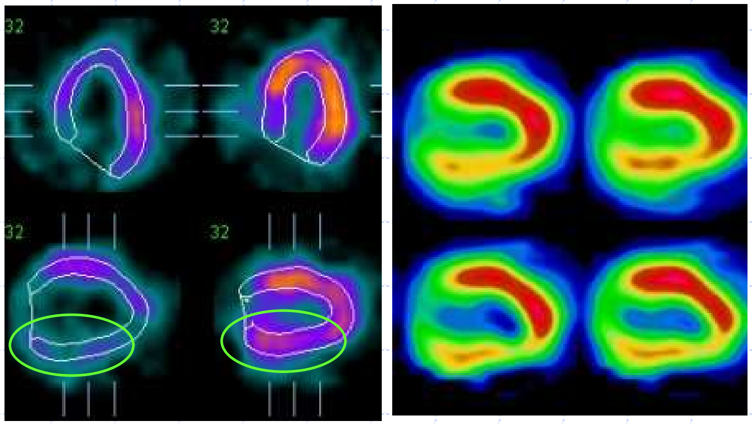
Pour une épaisseur diastolique de 15 mm

↳ 75 % de l'activité mesurée



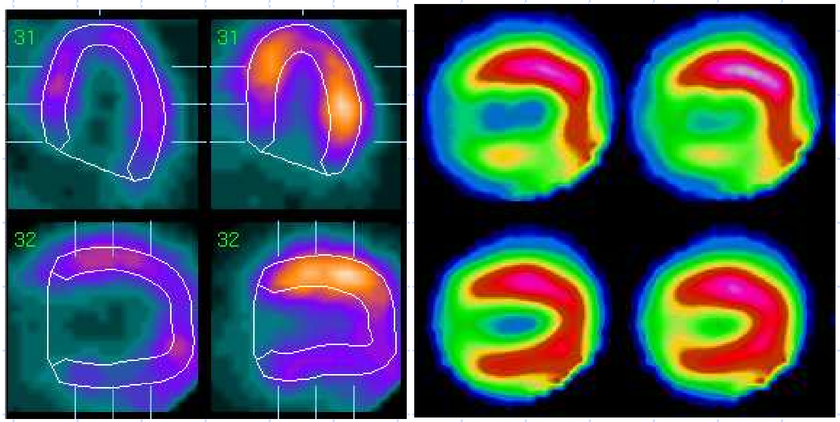
JR Galt et al. IEEE Trans Med Imaging 1990; 9:144-150

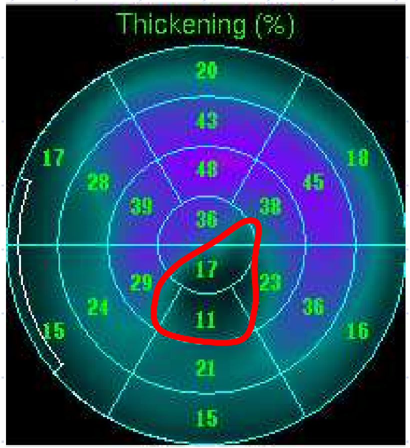
Applications cliniques



Examen normal

Se (%) p(T+/M)	NC	88
	ATT	86
Spé (%) p(T-/non M)	NC	50
	ATT	79

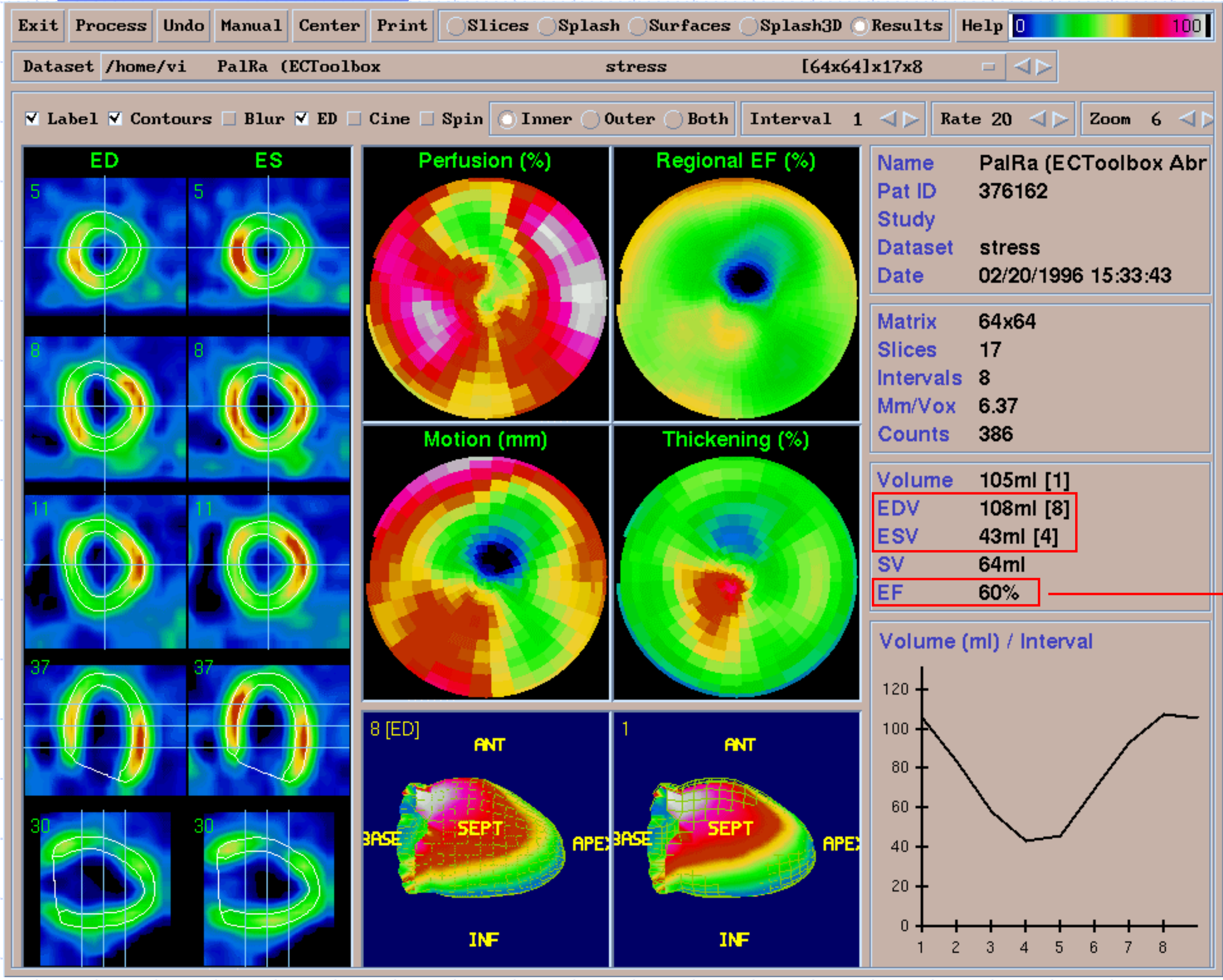


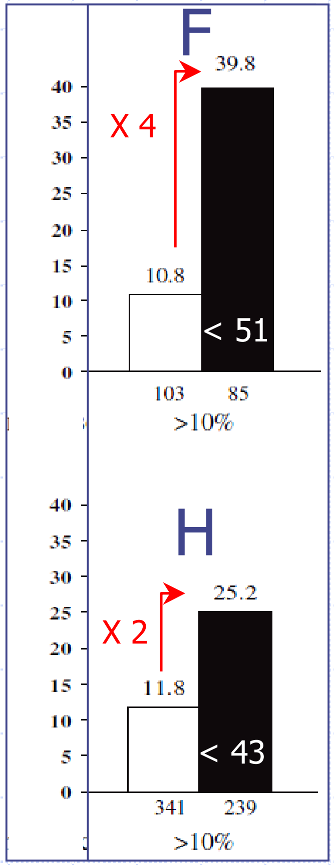


Sténose de l'artère circonflexe > 80%

J.M. Links J Nucl Cardiol 2002;9:183-187 - T.M. Bateman. Semin Nucl Med 2005;35:37-51 – G. Mowatt. Health Technology Assessment 2004; 8(30):1-222 (3032 patients)

Applications cliniques

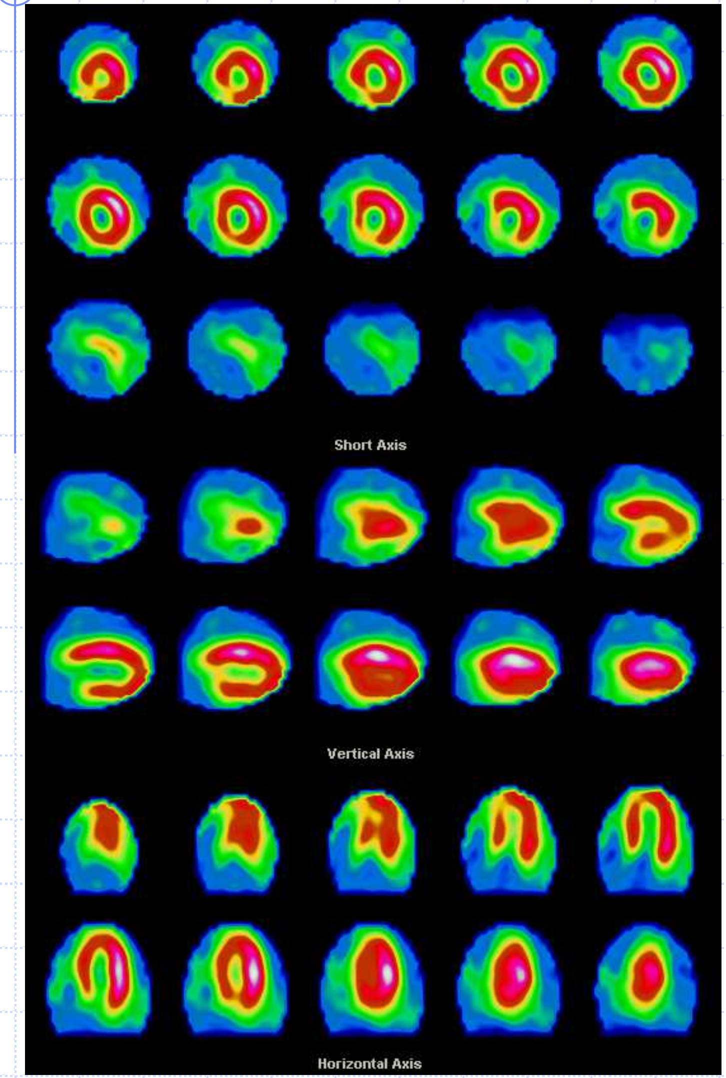


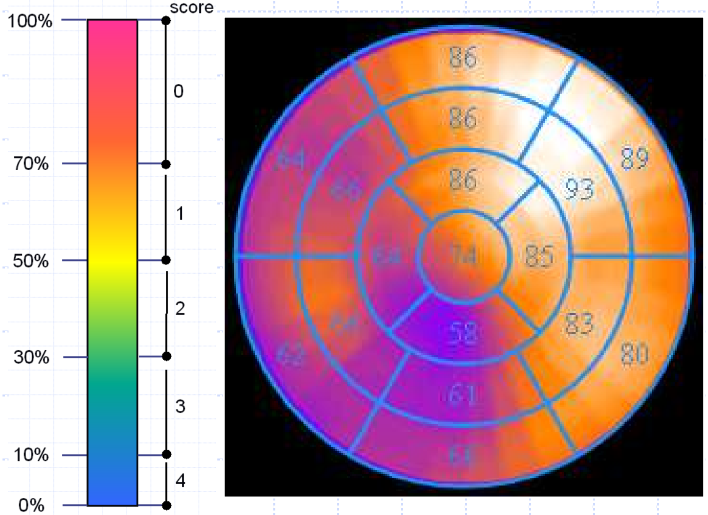


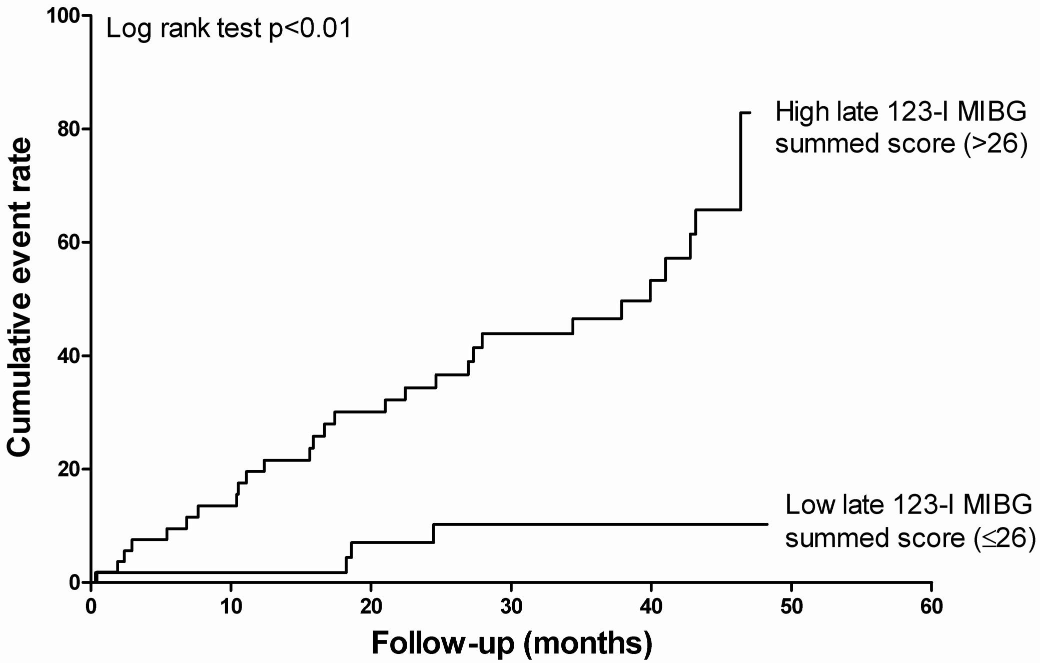
Evènements CV à 3 ans

T. Sharir et al. Circulation 1999;100:1035-42 et J Nucl cardiol 2006;13(4):495-506

MIBG



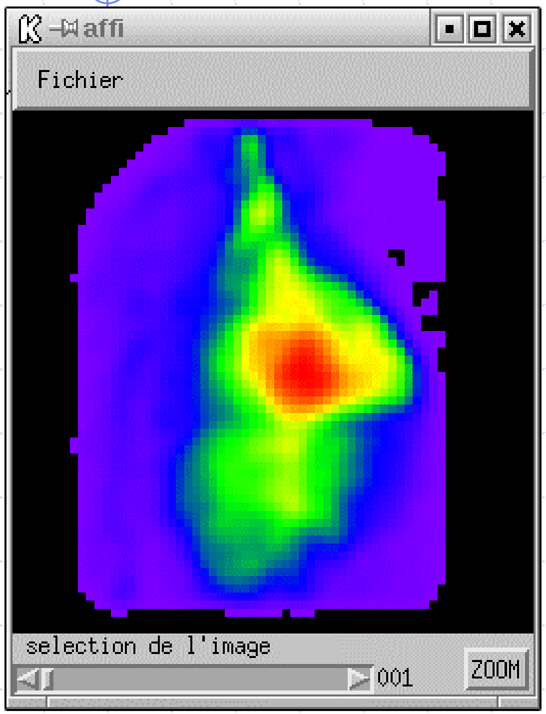


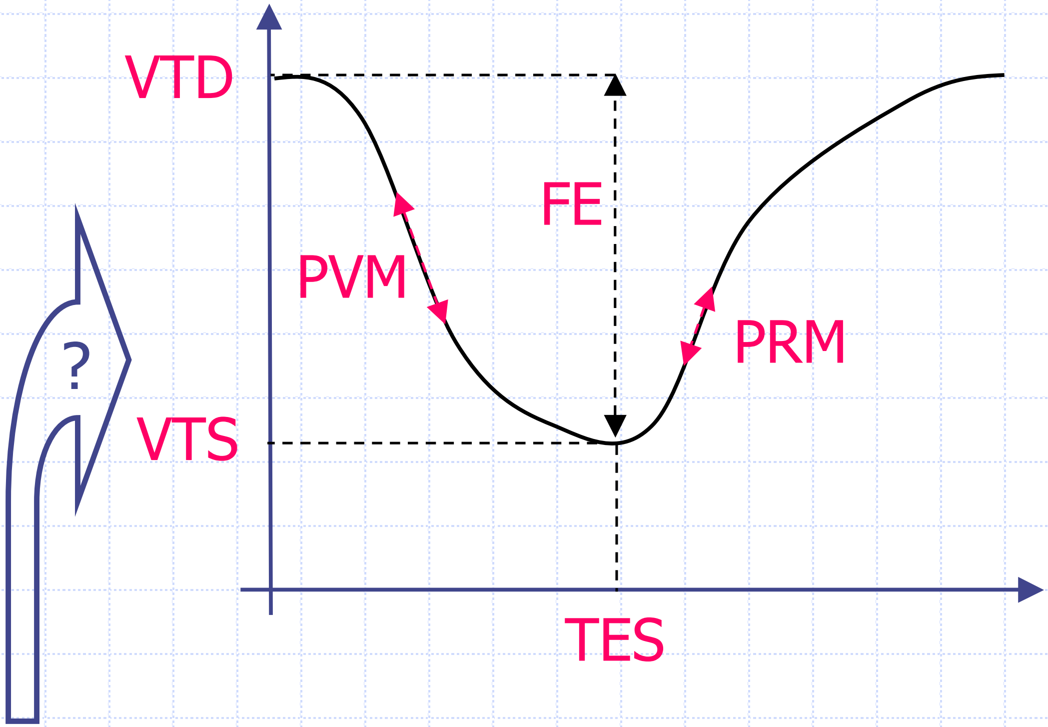


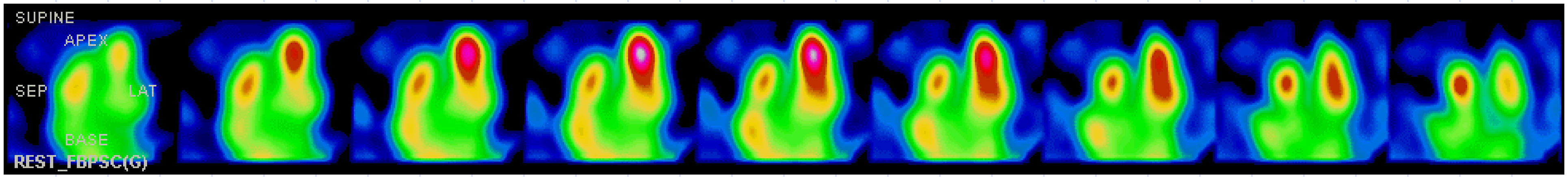
MJ Boogers et al. JACC 2010; 55(24):2769-2777

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)**

TOMO-VENTRICULOGRAPHIE







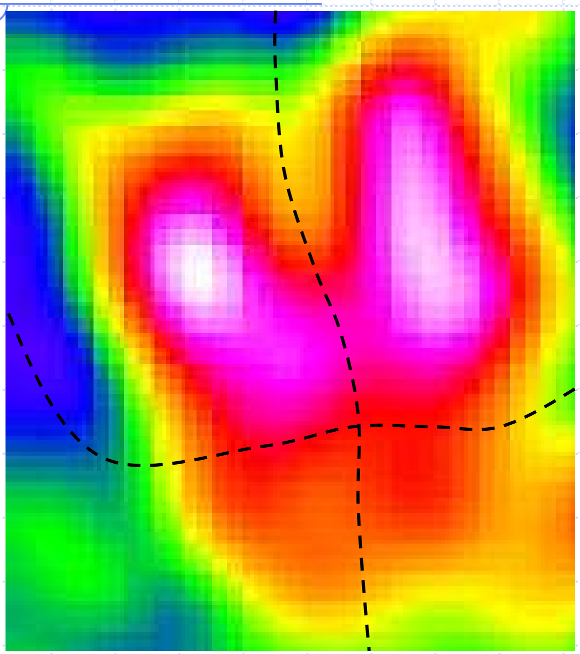
Choix d'un algorithme de segmentation

- Approche « frontières »
  - Seuillages (simple, d'histogramme, hystérésis…)
  - Dérivatives (gradient, Laplacien)
  - Surfaciques (Hueckel)
  - Morphologiques (gradient, ligne de partage des eaux)
  - Réseau de neurones
  - Champs de Markov
  - Variationnelles

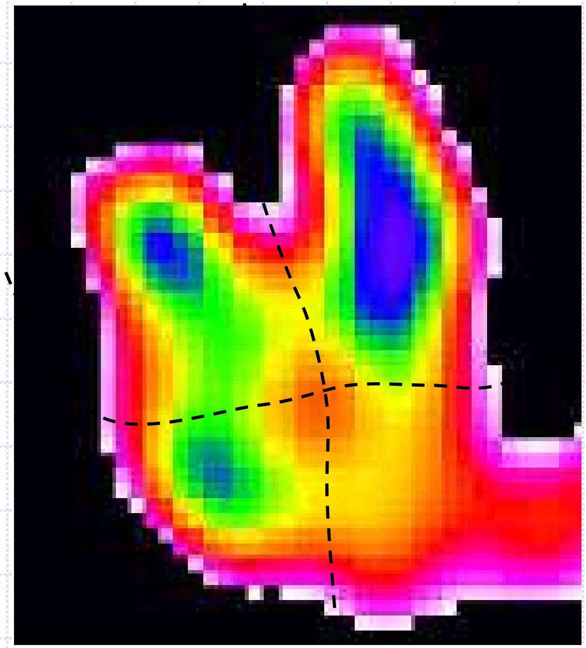
- Approche « régions »
  - Techniques Markoviennes,
  - Croissance de régions,
  - Division et fusion de graphes,
  - Etc.

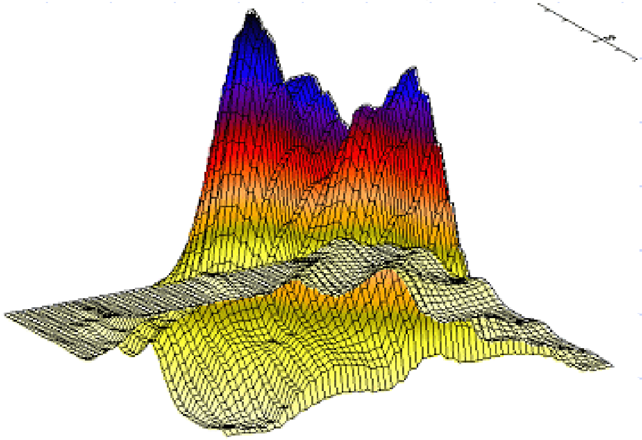
Analyse d'images: filtrage et segmentation", J.P Cocquerez et S. Philipp, 1995, MASSON.

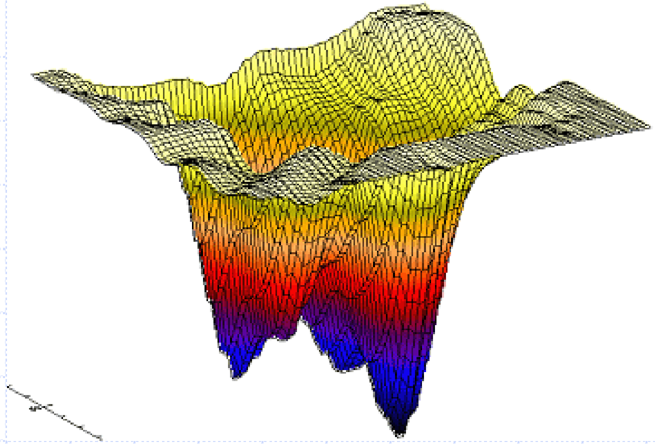
Segmentation



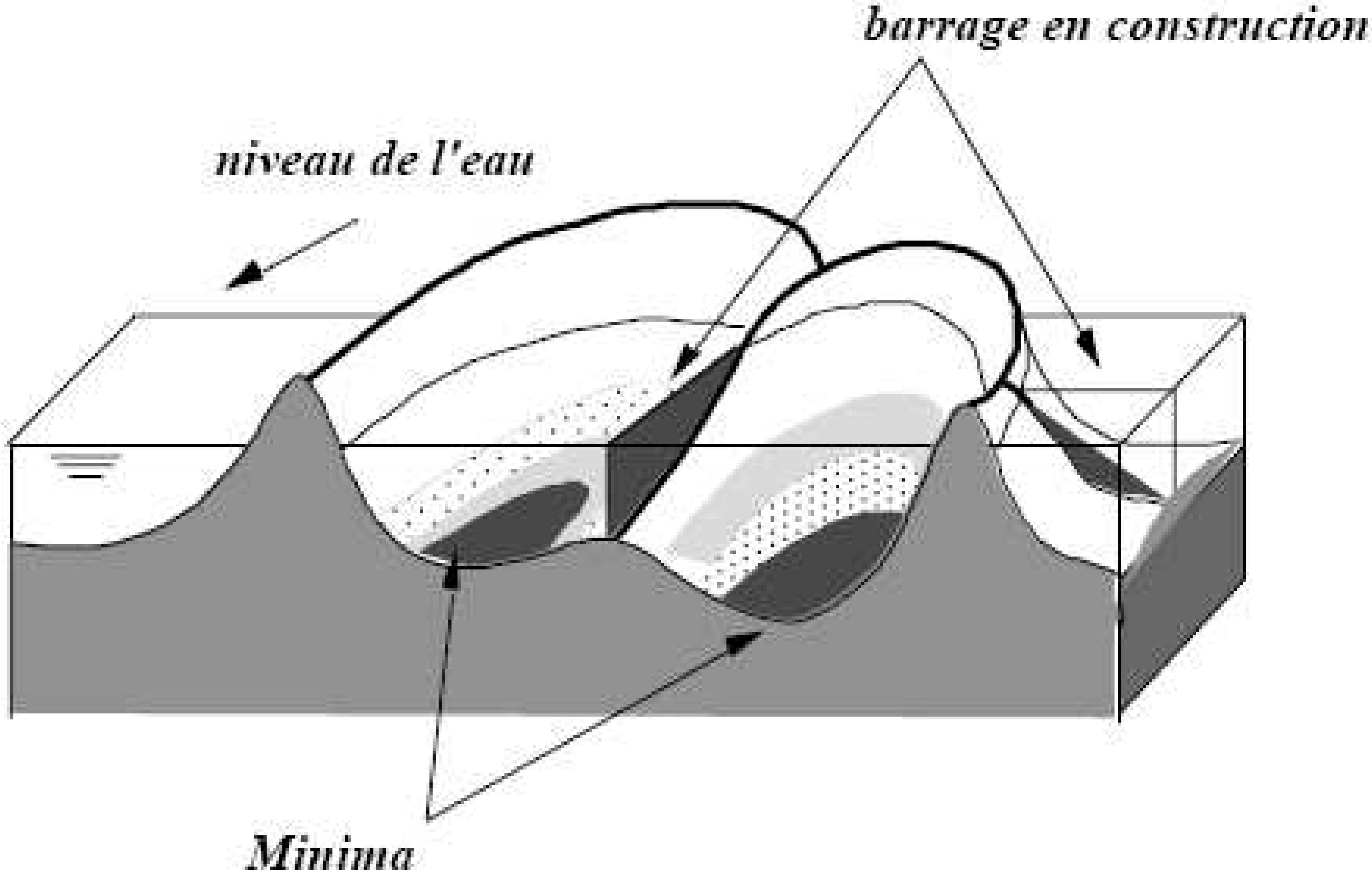
$$NG = NG_{max} - NG$$







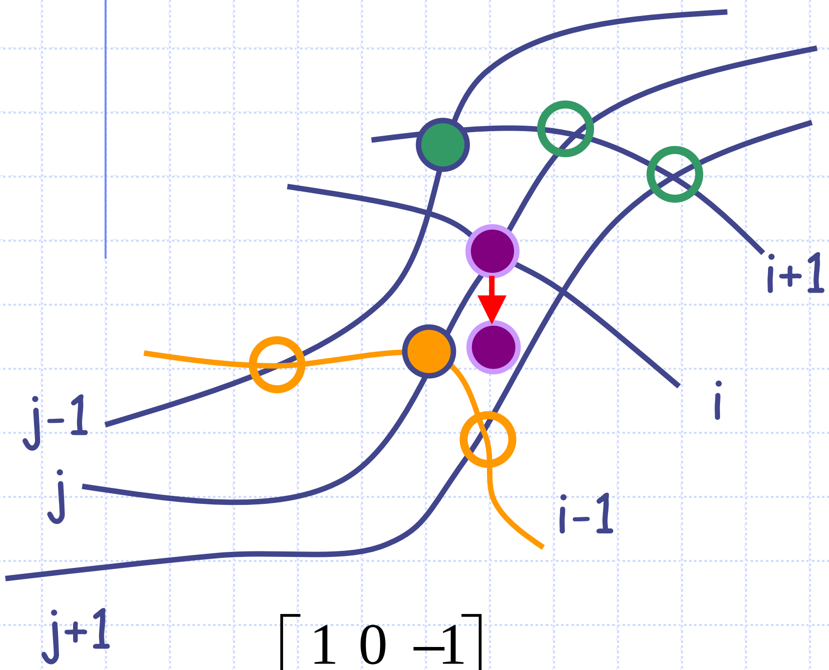
LPE par immersion



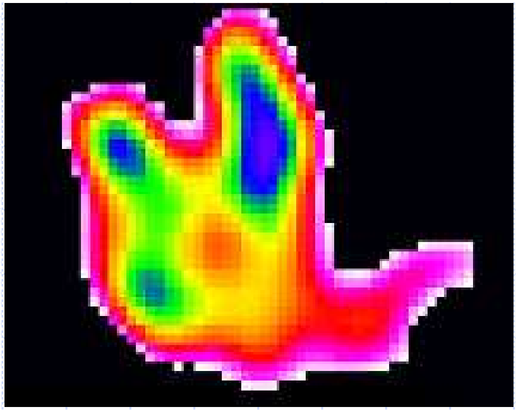
Schmitt M, Mattioli J. « Morphologie mathématique ». Paris, Masson, 1993.

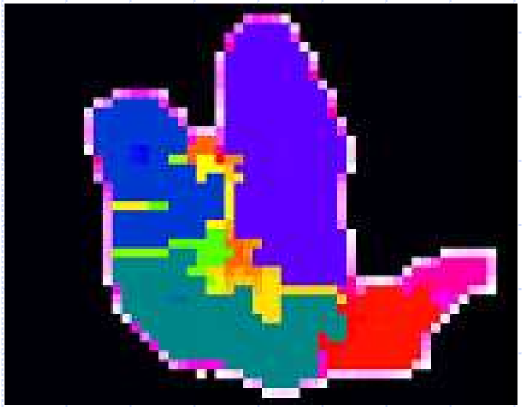
LPE par amincissement homotopique

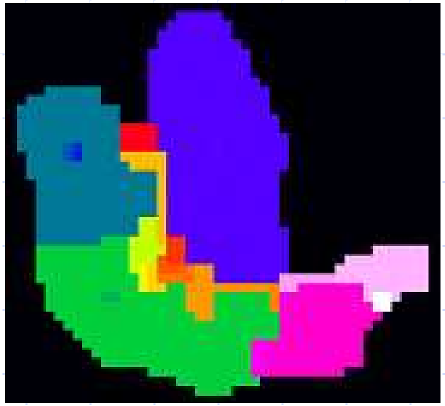
$$\text{si} \quad f_{max} < f(i,j) \le f_{min} \quad \text{alors} \quad f(i,j) = f_{max}$$



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



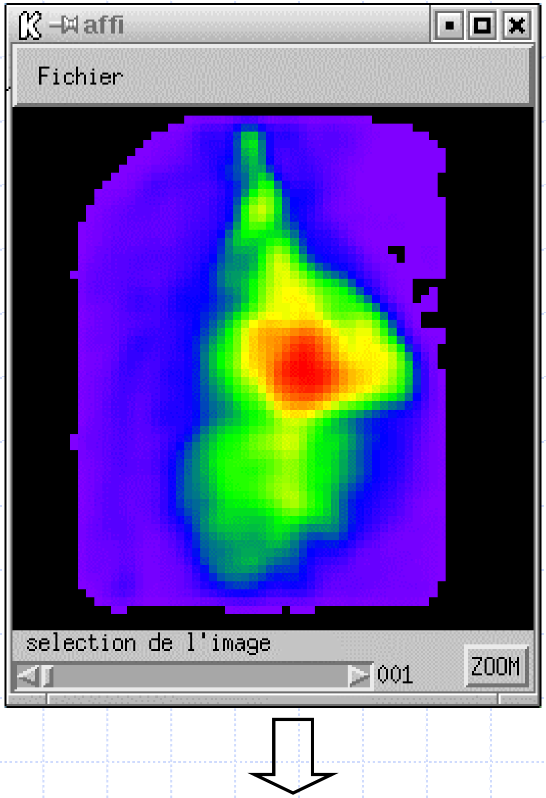


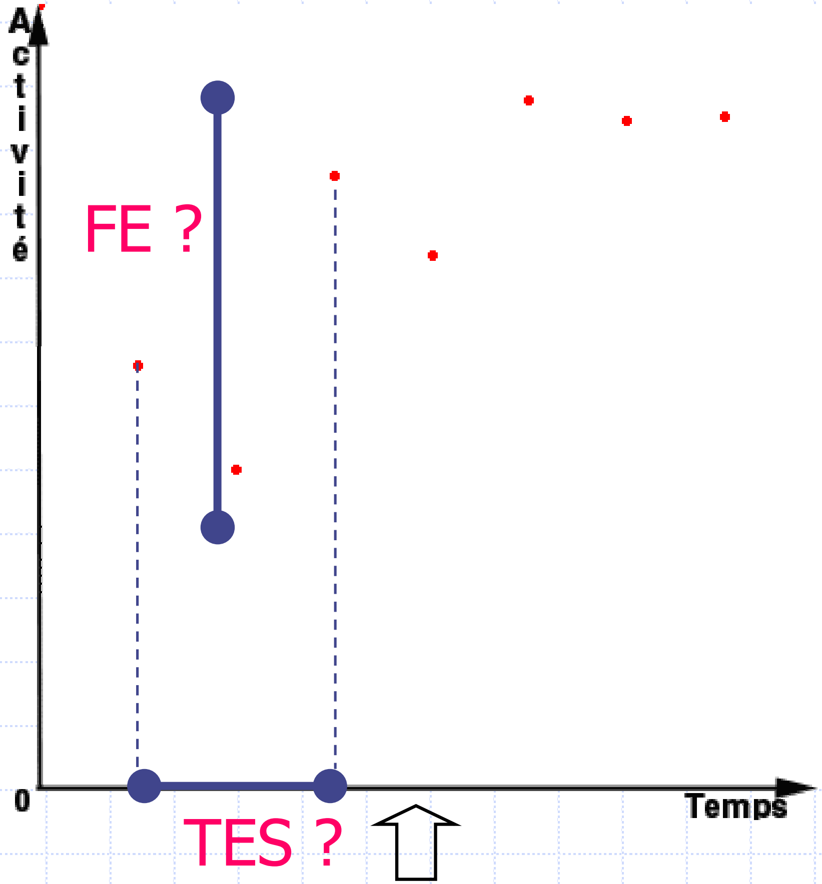


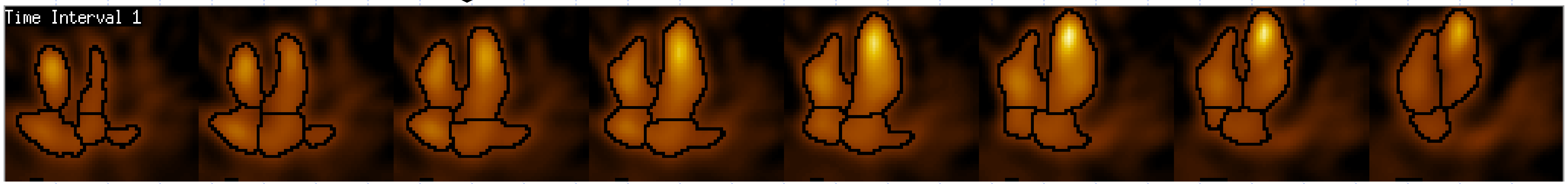
$$L' = \begin{bmatrix} 0 & 1 & 0 \\ 1 & -1 & 1 \\ 0 & 0 & 0 \end{bmatrix}$$

Mariano-Goulart et al.EJNM 1998; 22:1300-07 et Revue Acomen 2000;6:69-77

Extraction



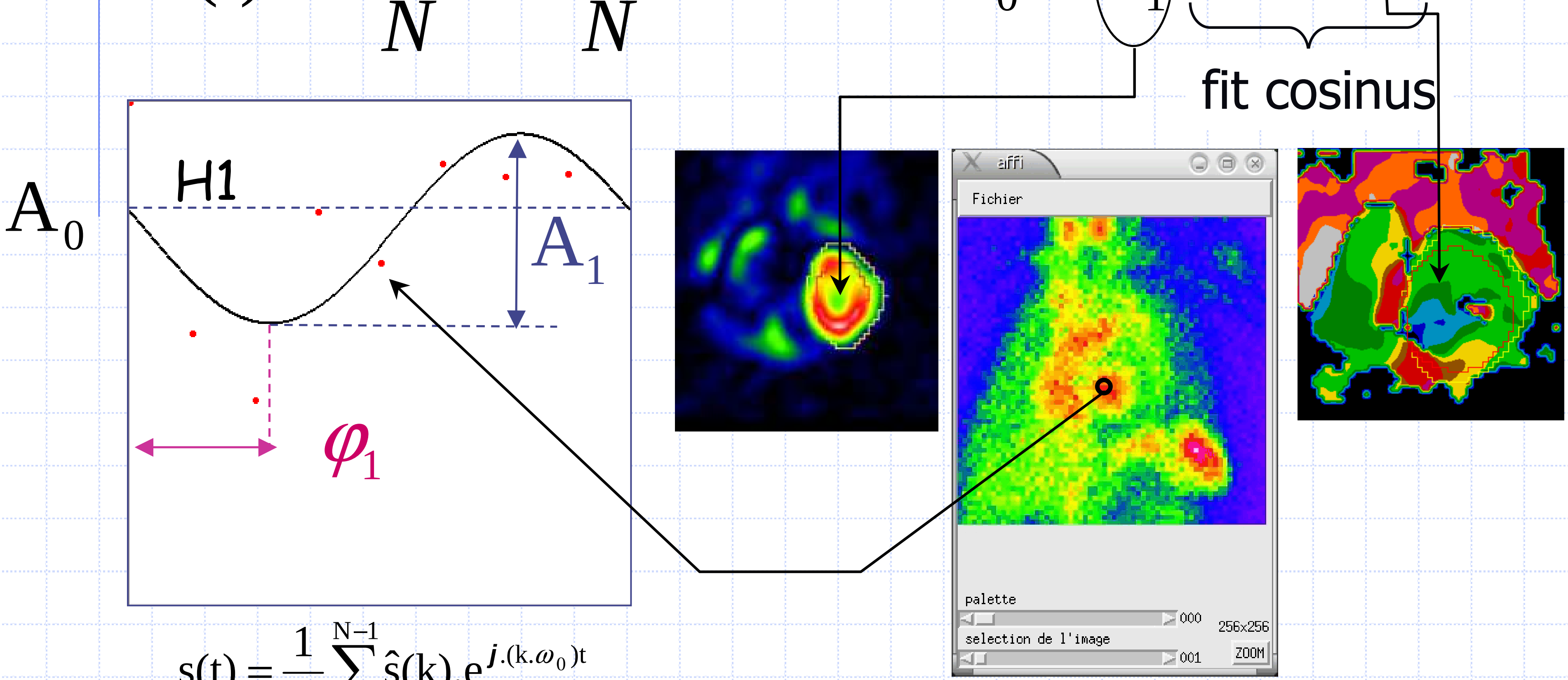




Mariano-Goulart et al.EJNM 1998;22 et EJNM 2001;28- Daou et al. JNM 2001;42

Ajustement mono- harmonique

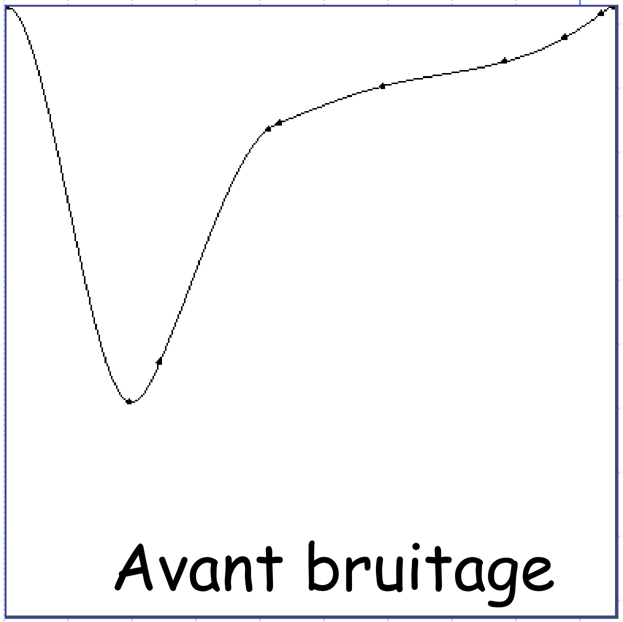
$$s(t) = \frac{\hat{s}(0)}{N} + \frac{\hat{s}(1)}{N} e^{j.(\omega_0)t} = A_0 + A_1 e^{j.(\omega_0 t + \varphi_1)}$$

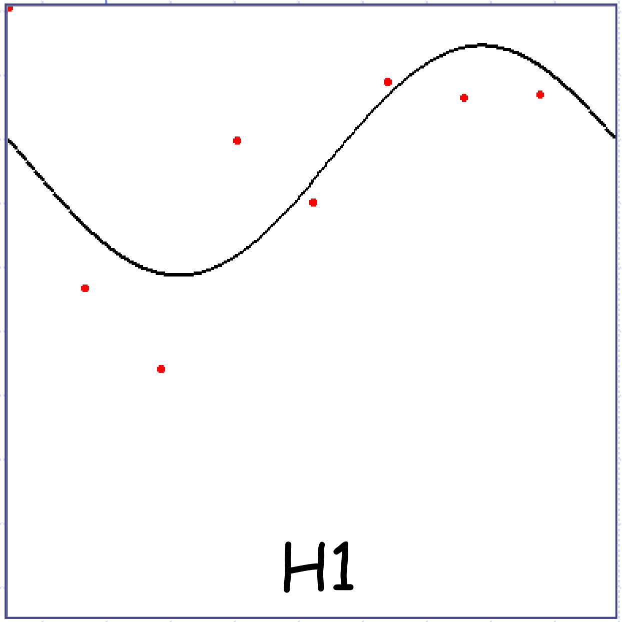


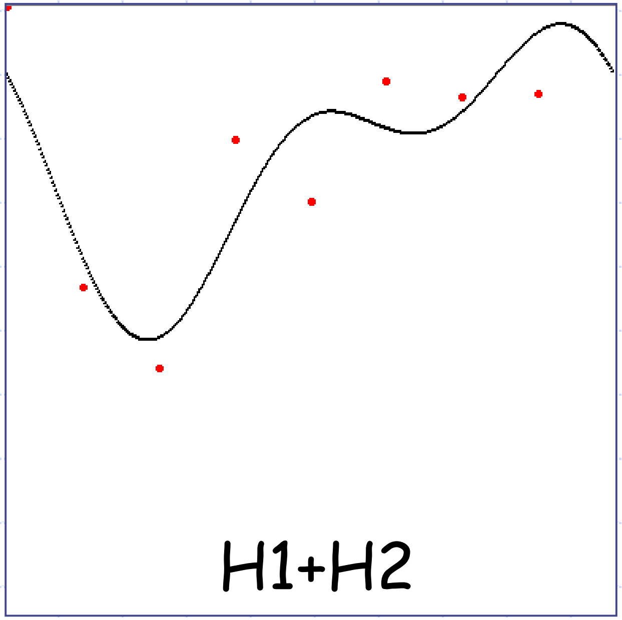
$$s(t) = \frac{1}{N} \sum_{k=0}^{N-1} \hat{s}(k).e^{j.(k.\omega_0)t}$$

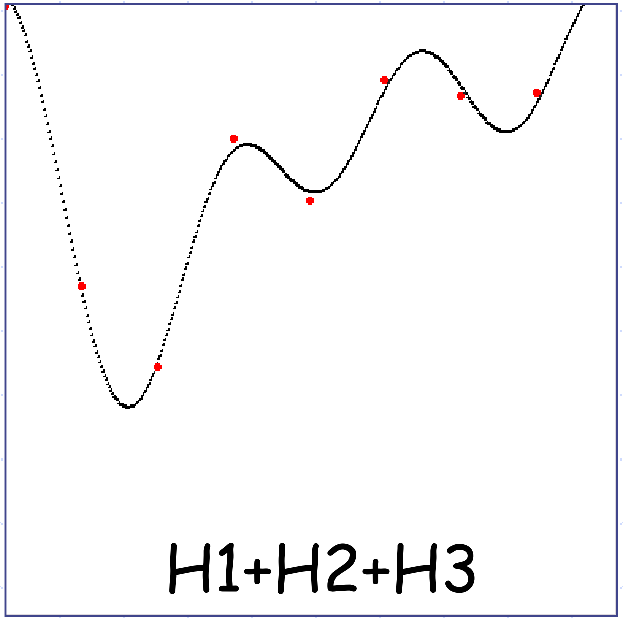
Ajustement multi-harmonique

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

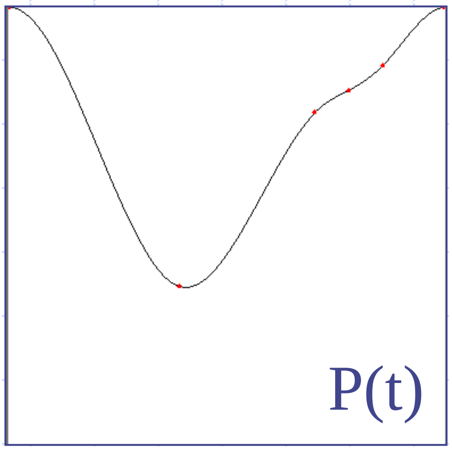


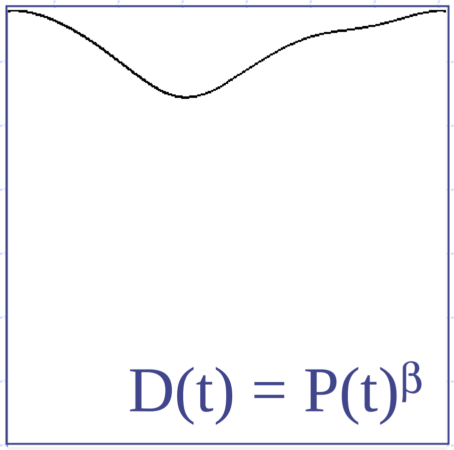


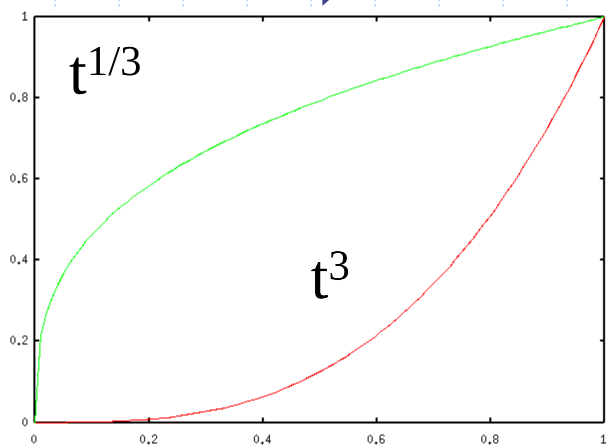


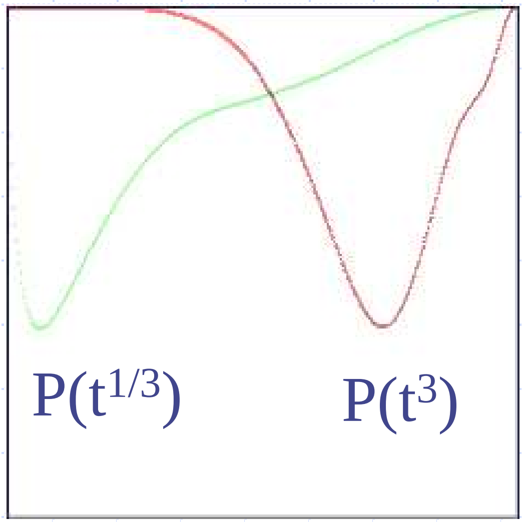


Déformation d’une CTA de référence

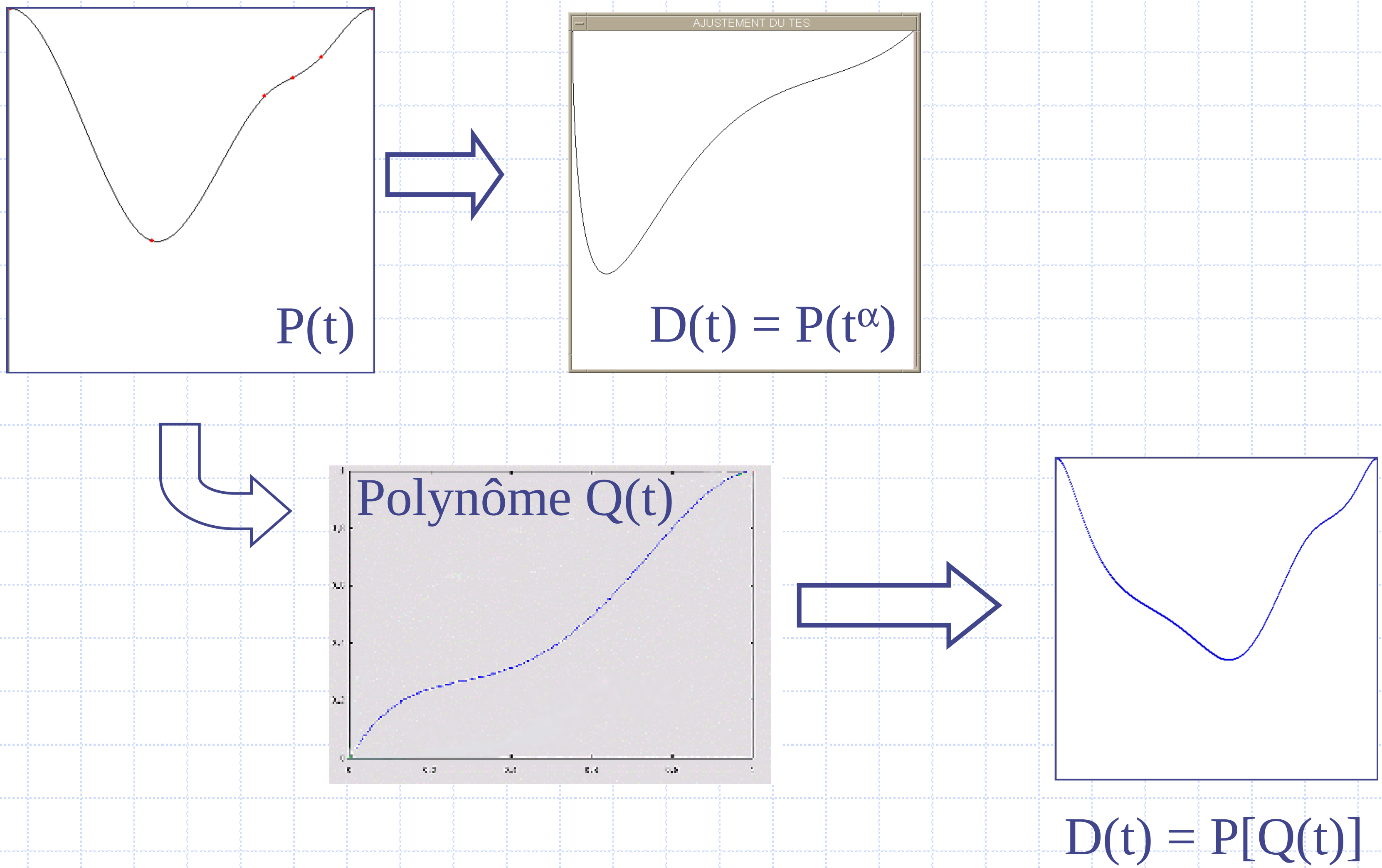




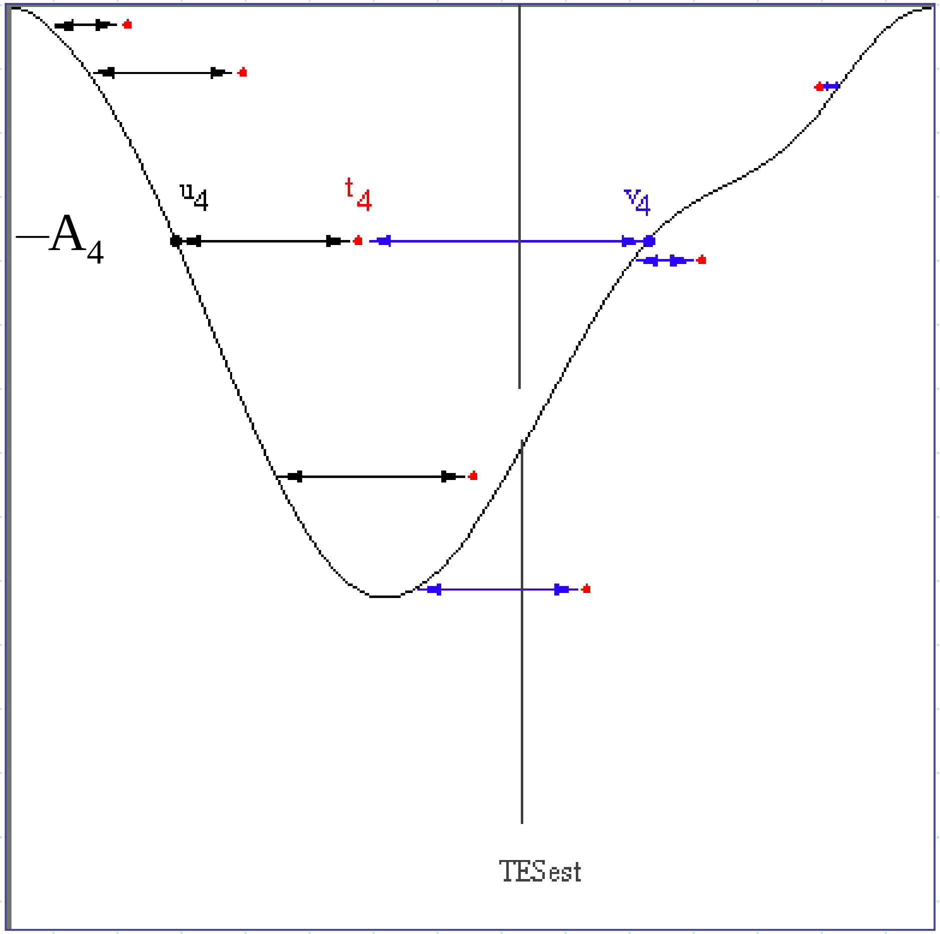




Déformation d’une CTA de référence

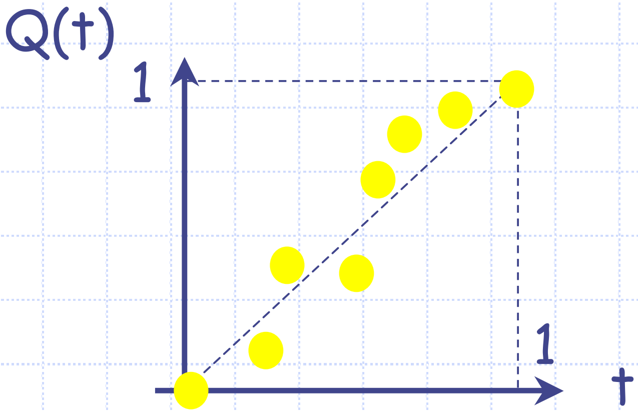


Détermination de Q(t)



Acquisition bruitée (t_4 , A_4)

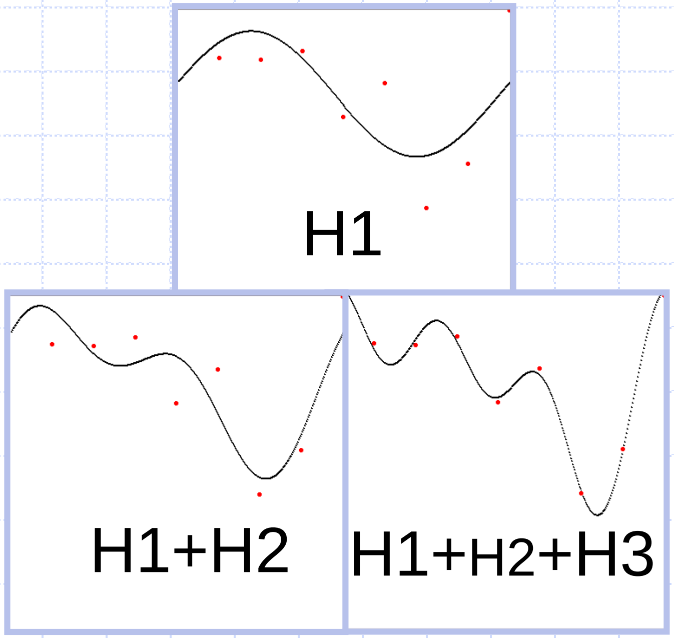
$$A_4 = P(u_4)$$
$$A_4 = D(t_4) = P[\,Q(t_4)]$$
$$\boxed{Q(t_4) = u_4}$$



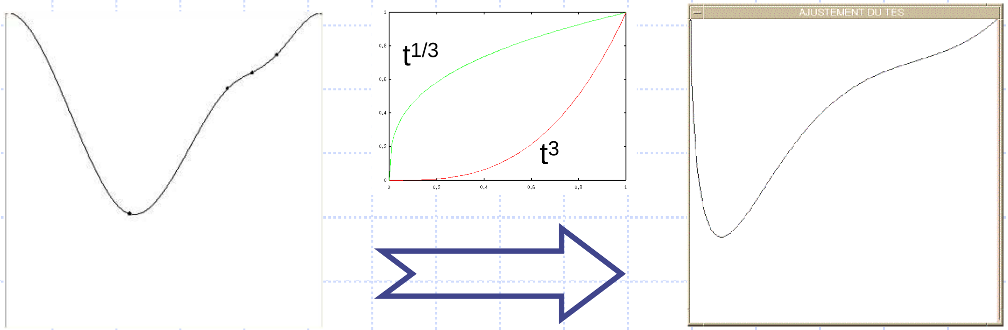
C Caderas de Kerleau & D Mariano-Goulart. IEEE Trans Med Imaging 2004 ; 23(4) : 485-91.

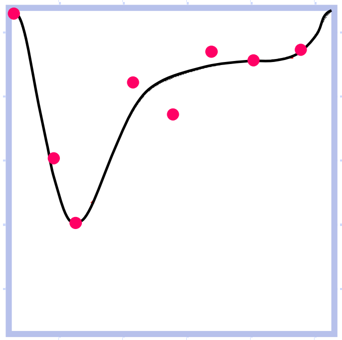
Validation

- Filtrage linéaire de la CTA
  - H1 : imprécis
  - Multi-harmoniques : bruité



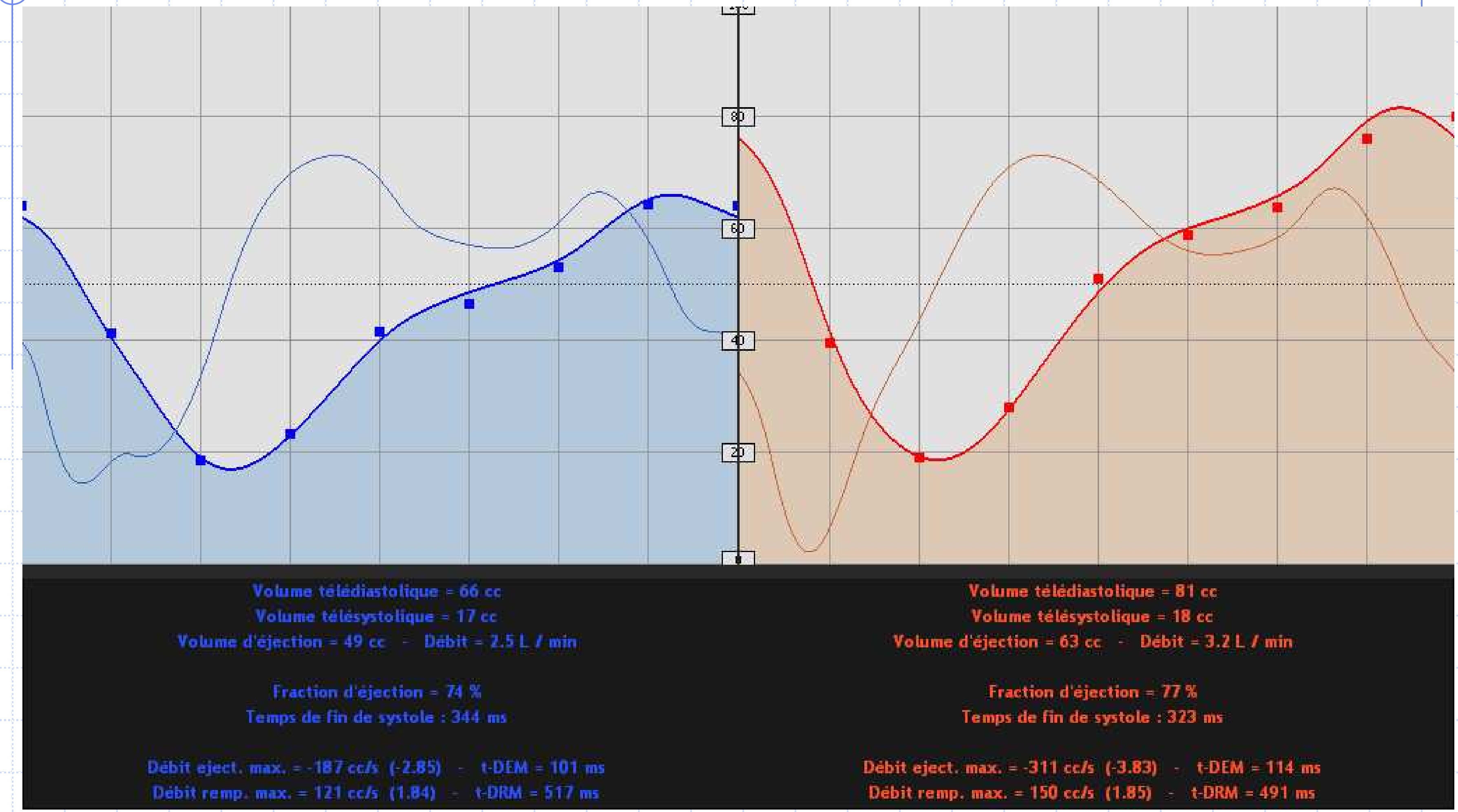
- Modélisation de la CTA



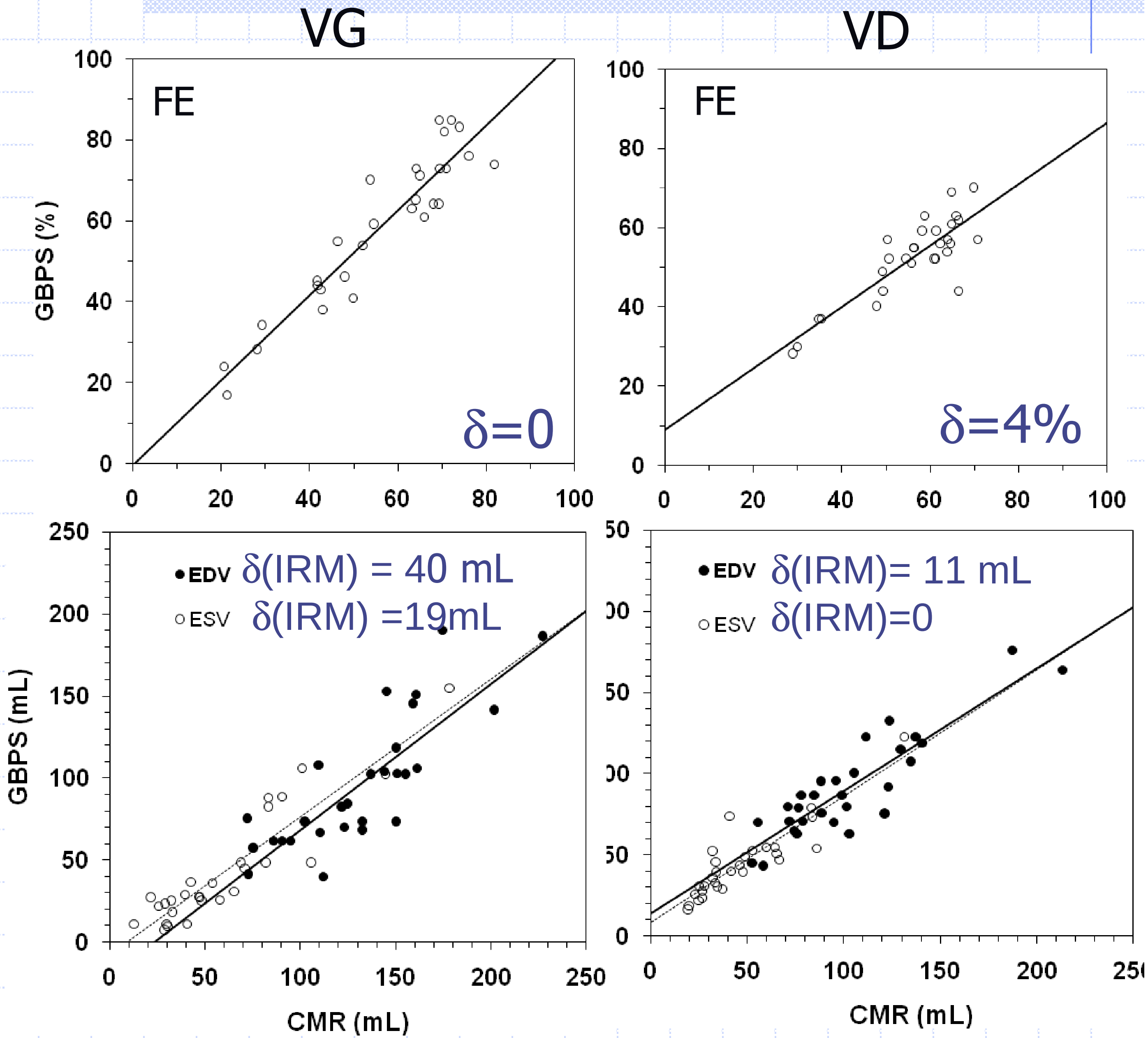


C Caderas de Kerleau & D Mariano-Goulart. IEEE Trans Med Imaging 2004 ; 23(4) : 485-91.

Analyse de la fonction systolique



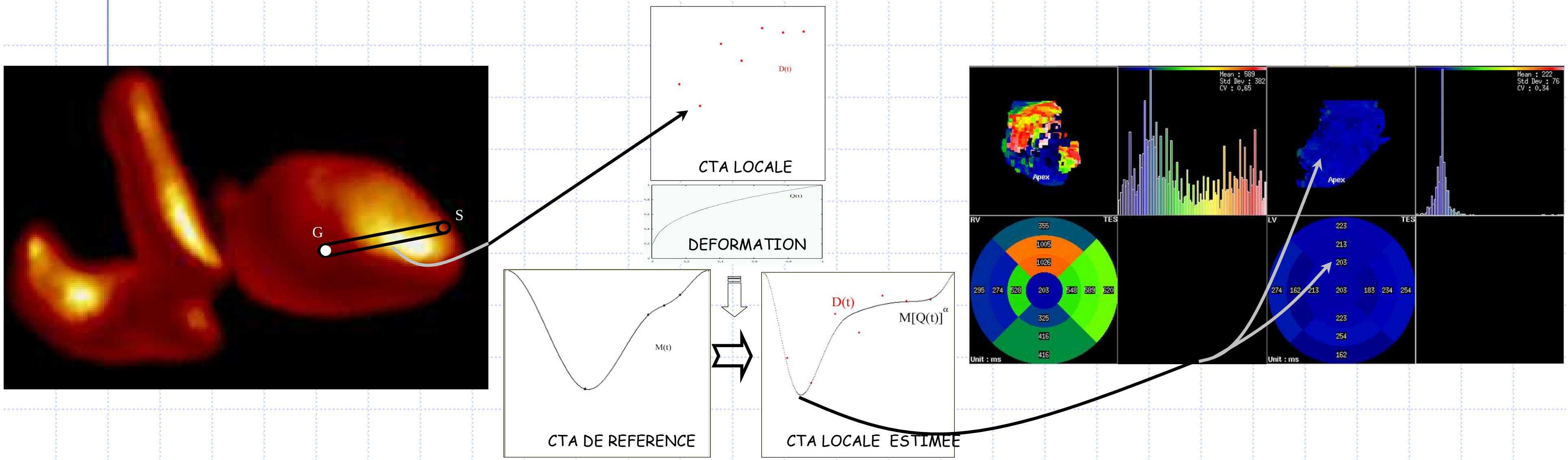
TVI versus IRM



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

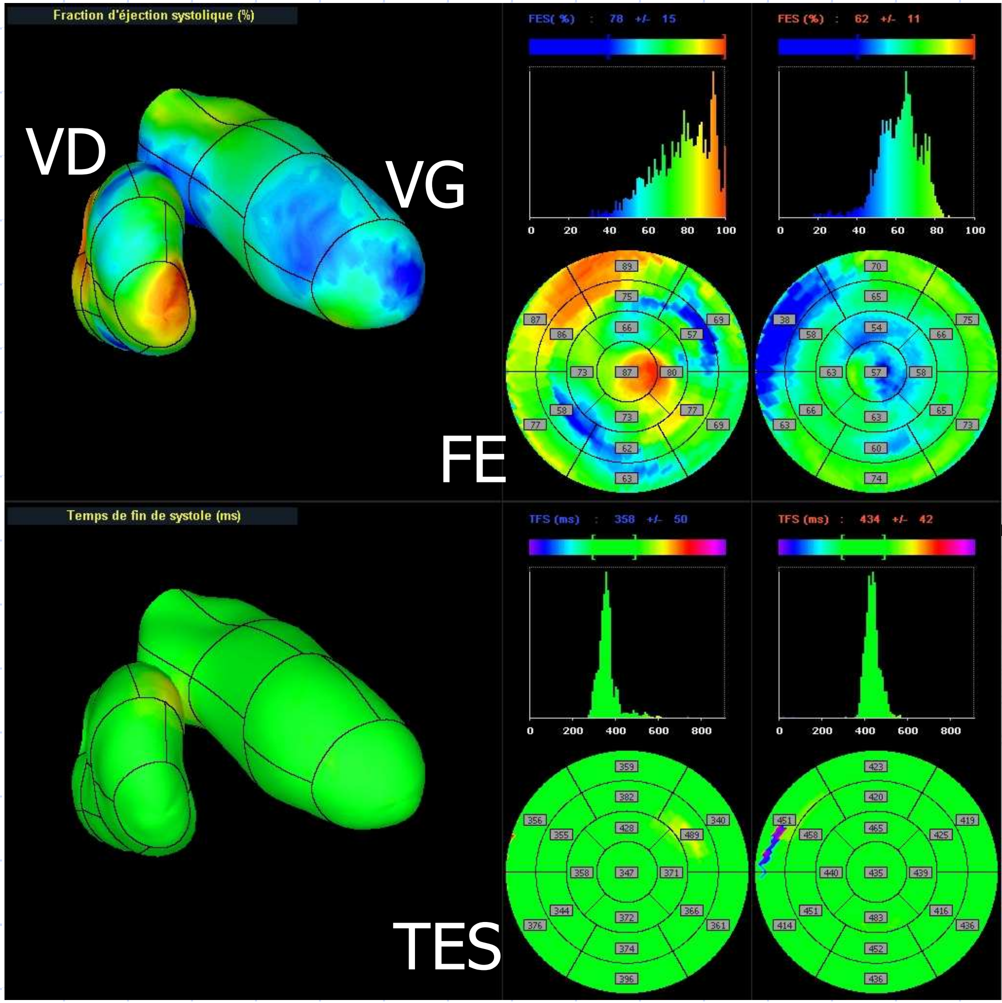
L. Sibille et D. Mariano-Goulart, An. Nucl. Med 2010.

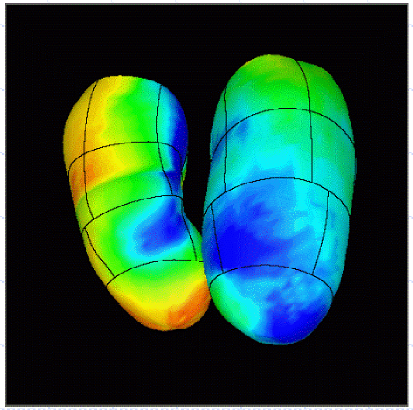
ANALYSE 3D DE CTA LOCALES



Mariano-Goulart et al. J Nucl Med. 2007; 47: 1416-1423

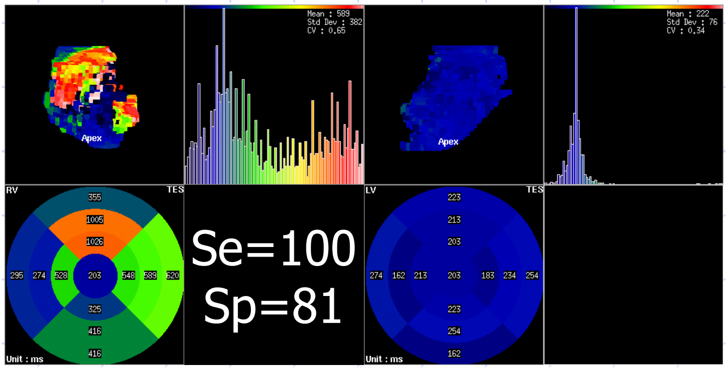
ANALYSE 3D DE CTA LOCALES





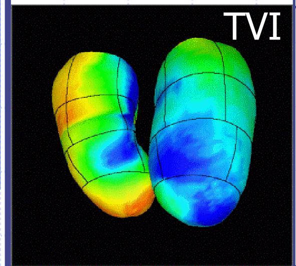
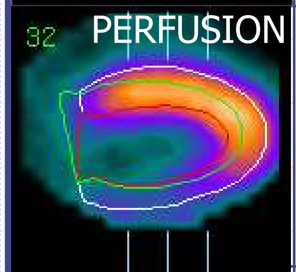
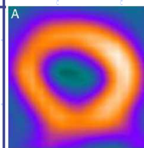
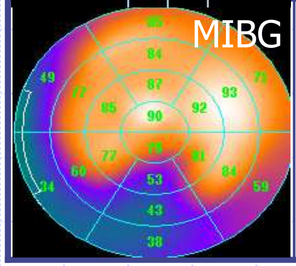
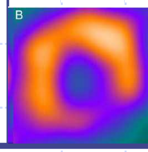
Parameter	Control subjects	Patients with localized ARVD	Patients with diffuse ARVD
EF (%)	63 ± 7	60 ± 5	33 ± 12†
EDV (mL)	101 ± 17	107 ± 20	180 ± 81‡
ESV (mL)	37 ± 8	43 ± 10	124 ± 70†
σ-EF (%)	13 ± 3	14 ± 2	11 ± 3
σ-TES (ms)	63 ± 24	167 ± 64†	277 ± 106†

S=80 ms



Mariano-Goulart et al. J Nucl Med. 2007; 47: 1416-1423

Synthèse et résultats

Modalité	Logiciels Déconvolution Reconstruction +	FE	Volumes	Cicatrice	Hypo-kinésie	Dys-kinésie
	LPE Fit de CTA Échantillonnage	+	+		+	+
	Identification Modélisation Quantification	Pronostic		 +	+	+
	Identification Modélisation Quantification	-	-	 +	-	-

Pour aller plus loin…

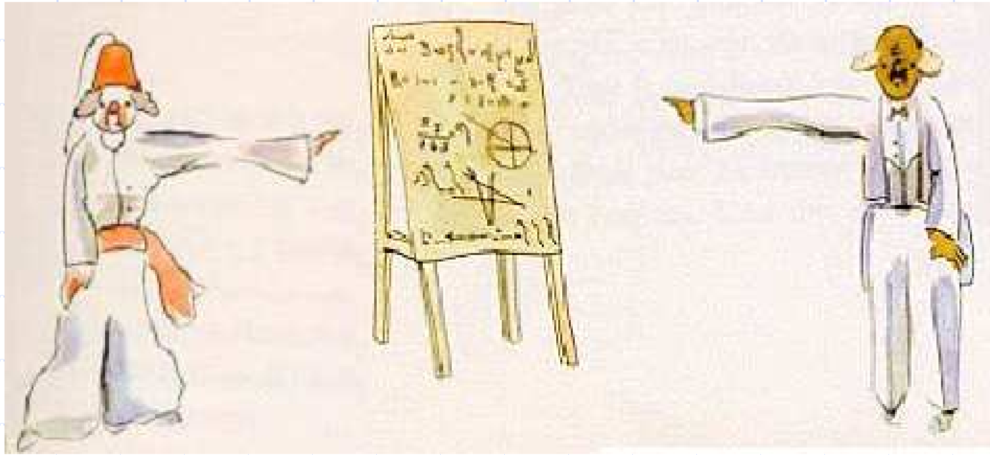
- ✓ The Mathematics of Computerized Tomography.
F. Natterer. 2001. SIAM.

- ✓ Quantitative Analysis in Nuclear Medicine Imaging.
H. Zaidi Ed. Springer.
(http://www.springerlink.com/content/k513k6/?p=be16e74068244dbfbeb4d7123ee136b3&pi=0)

- ✓ 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 pour votre attention…

d-mariano_goulart@chu-montpellier.fr
http:\\scinti.etud.univ-montp1.fr