ACIMED
Imagerie Cardiovasculaire

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Plan

1. Introduction aux maladies Cardiovasculaires.
2. Modalités d'imagerie, Modes d'imagerie.
3. Applications Cliniques.

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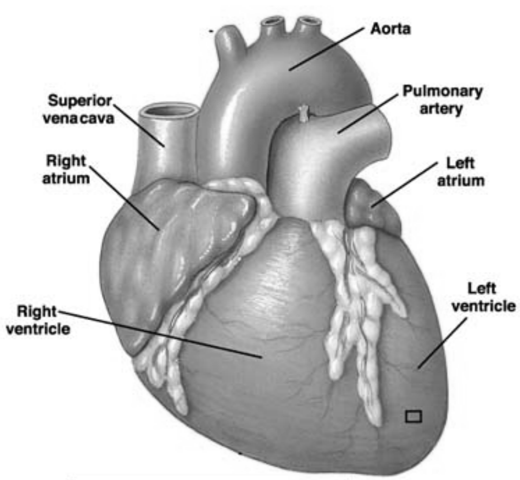
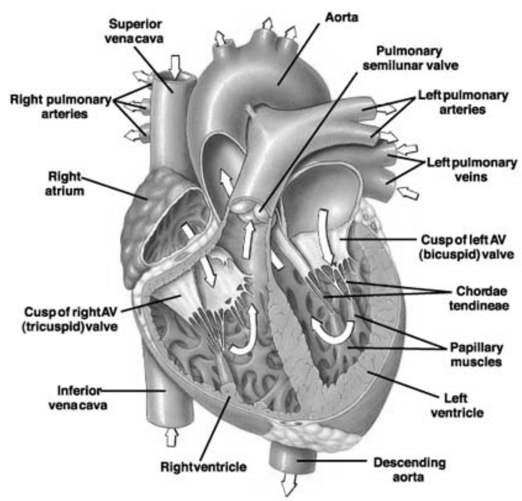
Introduction

Système Physiologique Cardiovasculaire

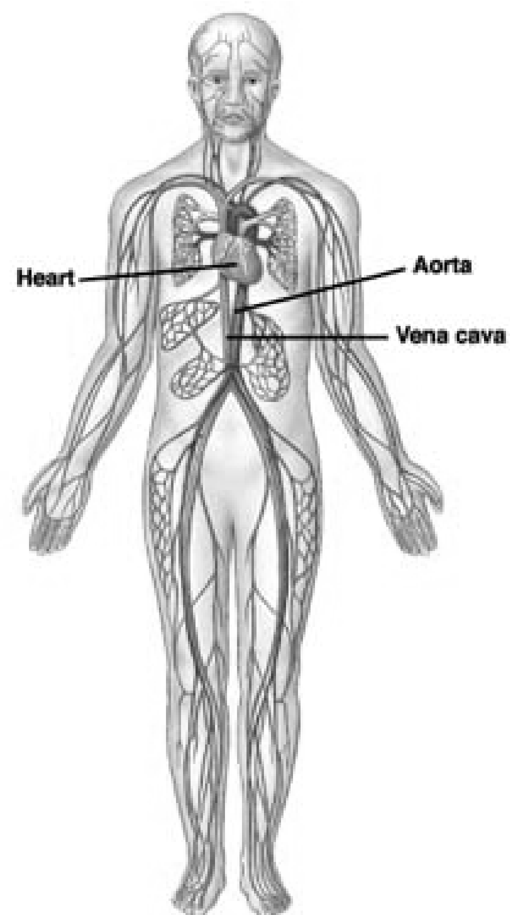
- Cardiaque
  - Ventricules, valves, myocarde
- Vasculaire
  - Artères et veines



The ventricles occupy the bulk of the heart. The arteries and veins all attach to the base of the heart.



One-way flow through the heart is ensured by two sets of valves.

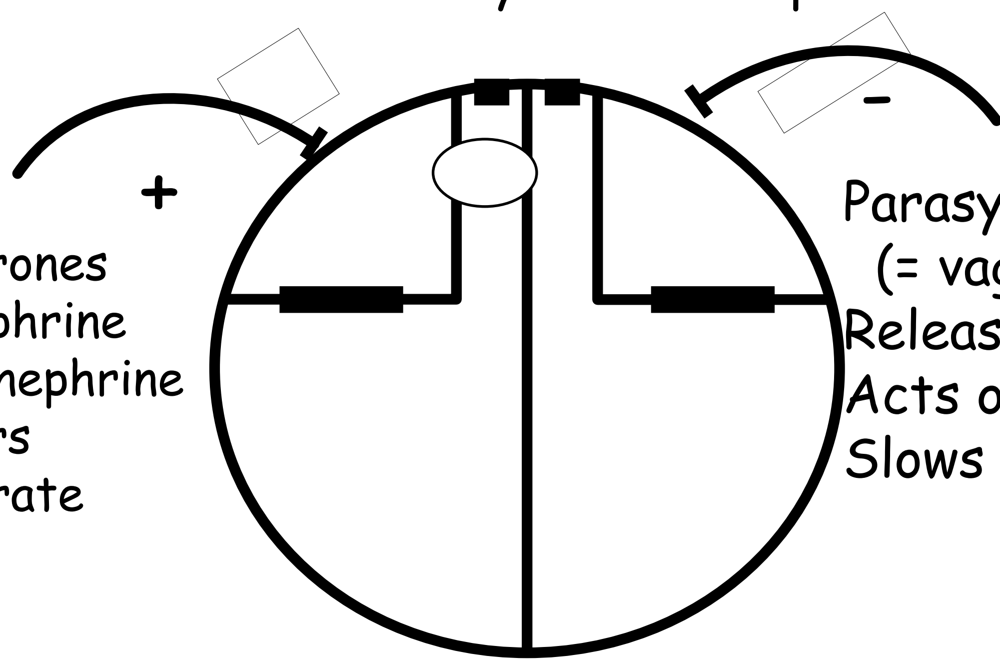


Introduction

Rythme Cardiaque

Rythmicité inhérente (~100bpm)
Démarre dans le SA node

Coeur est restreint par le système **vagal**
Rythme au repos ~70bpm



Sympathetic neurones
Release norepinephrine
+ circulating epinephrine
Act on β receptors
Increases heart rate

Parasympathetic neurones
(= vagus)
Releases ACh
Acts on muscarinic receps
Slows heart rate

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Introduction

Les Vaisseaux

	Ventricule	Chambre de pompage Haute pression
Grand diamètre paroi epaisse	Artères	**Elastiques**: elastin, régulation de la pression artérielle
Petit diamètre paroi epaisse	Arterioles	**Musculaires**: muscle mou Resistance régulation du flux dansle corps
Petit diamètre paroi fine	Capillaires	Echanges
Grand diamètre paroi fine	Vénules/veines	**Capacitance**
	Atrium	Chambre de stockage Basse pression

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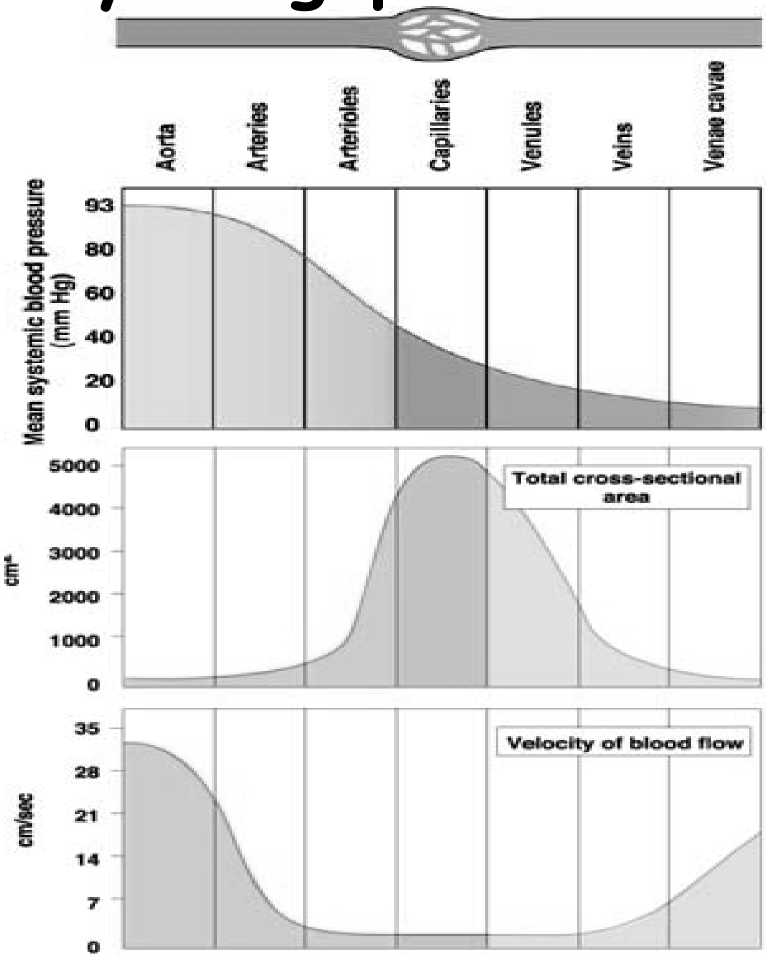
Introduction

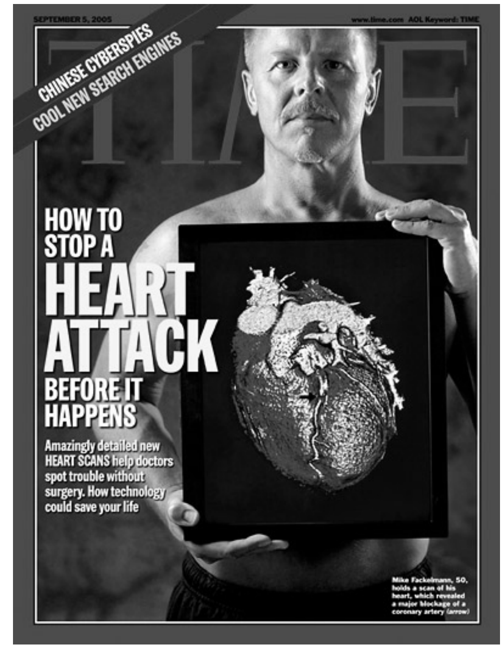
Système Physiologique Cardiovasculaire



Introduction

Incidence et Santé Publique des malaldies cardiovasculaires

- 1/3 des examens d'imagerie médicale dans le monde.
- Coût:
  - US (U),
  - CT-SPECT (3U),
  - IRM (5U),
  - PET(14U),
  - Cathérisation (20U)



2005

[Eugenio Picano, « Economic and biological costs of cardiac imaging »,*cardiovascular Ultrasound*, 3:13, 2005]

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Introduction

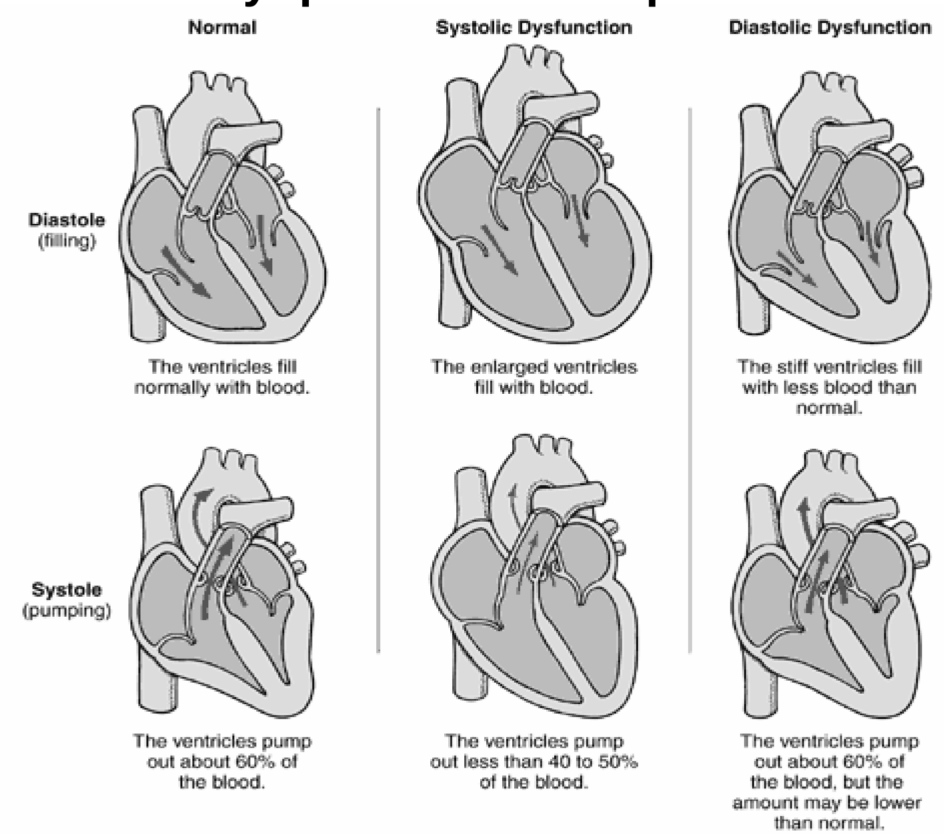
Pathologies

- Liste des Pathologies
  - Abnormal Heart Rhythms
  - Atherosclerosis
  - Cardiomyopathy
  - Infective Endocarditis
  - Pericardial Disease
  - Heart Valve Disorders
  - Heart Failure
  - High or Low Blood Pressure
  - Coronary Artery Disease
  - Peripheral Arterial Disease
  - Venous Disorders
  - Aneurysms and aortic dissection

Introduction

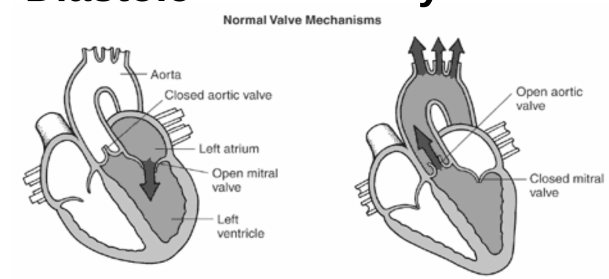
Pathologies

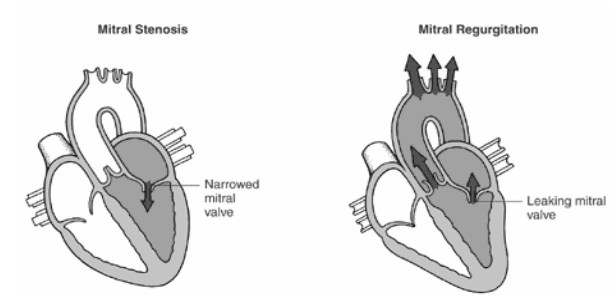
Myopathies Cardiaques



Myopathies des Valves

Diastole **Systole**

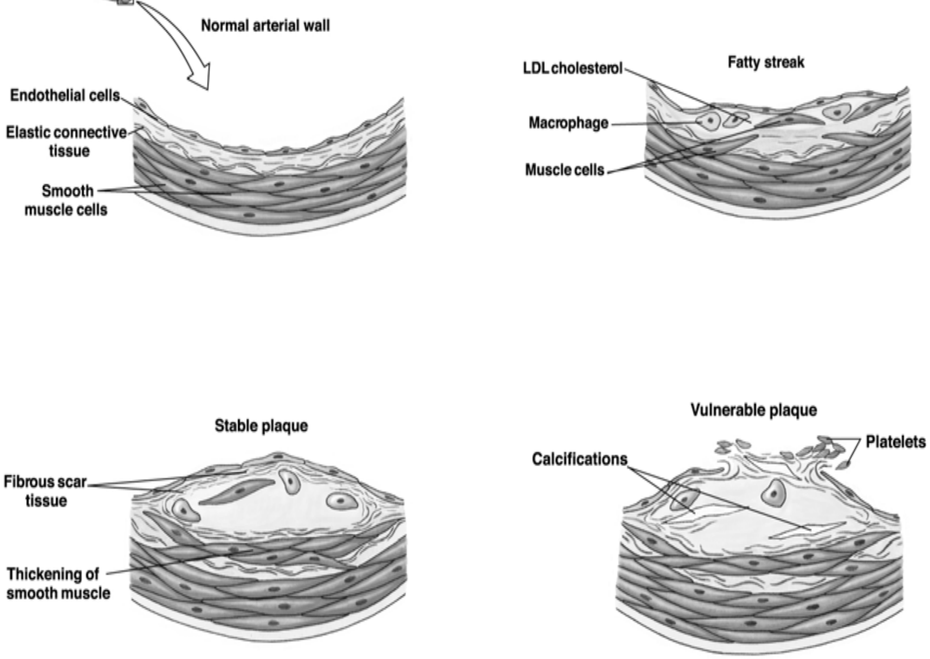




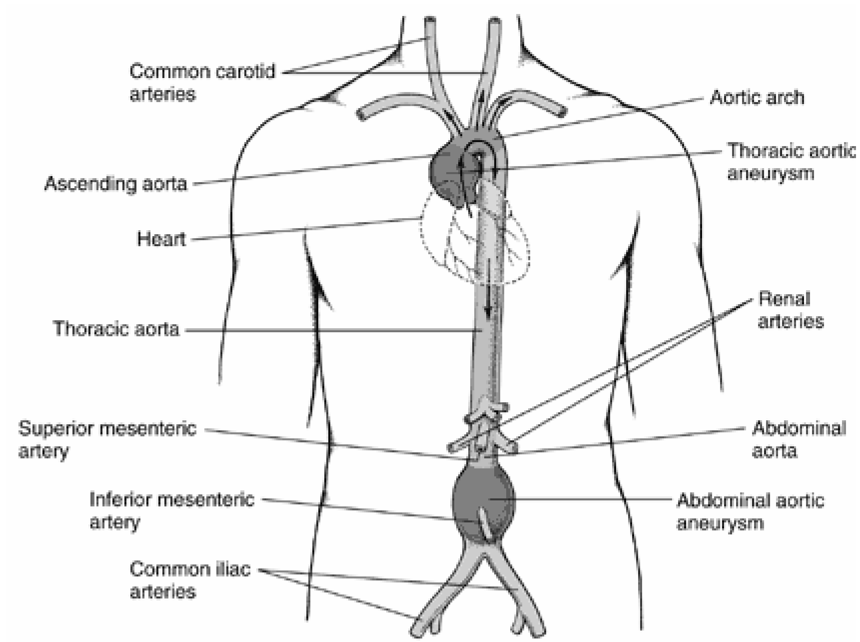
Introduction

Pathologies:

Plaques sur parois des artères



Anévrismes



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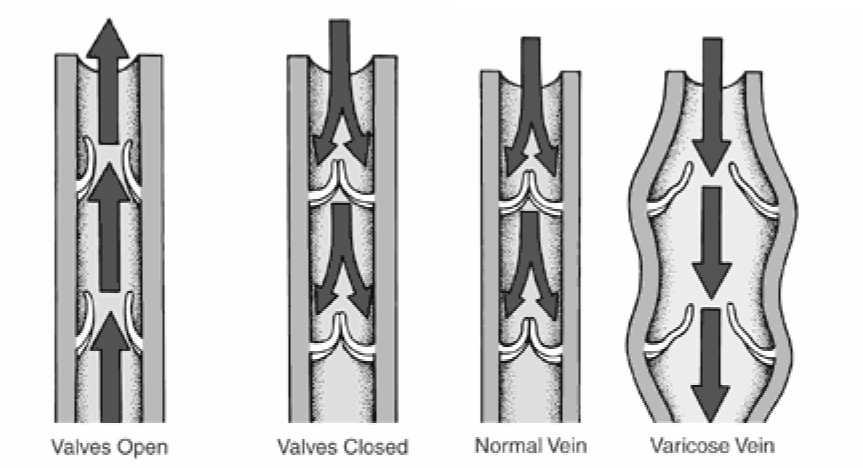
Introduction

Veines varisqueuses



Introduction

- Vaisseaux: Eléments pour le diagnostique
  - **Plaques**
    - Composition de la paroi des vaisseaux et des plaques.
    - Vulnérabilité des plaques.
    - Effets potentiels d'une intervention chirurgicale.
    - Effets de traitements médicamenteux.
  - **Anévrisme-thrombose**
    - Age.

Introduction

Imagerie Cardiovasculaire

- Modalités d'Imagerie:
  - X-Ray, CT, IRM, US,
  - PET-SPECT.
- Protocoles d'imagerie:
  - Repos
  - Dynamique
  - Effort (stress)
  - Perfusion
  - Agent de contraste

Philips Medical Systems

Plan

2.1 Modalités d'Imagerie

- X-Ray
- CT
- Ultrasons
- IRM
- PET/SPECT

2.2 Modes d'Imagerie

- Statique
- Dynamique
- Effort (stress exams): perfusion/oxygène
- Agents de contraste

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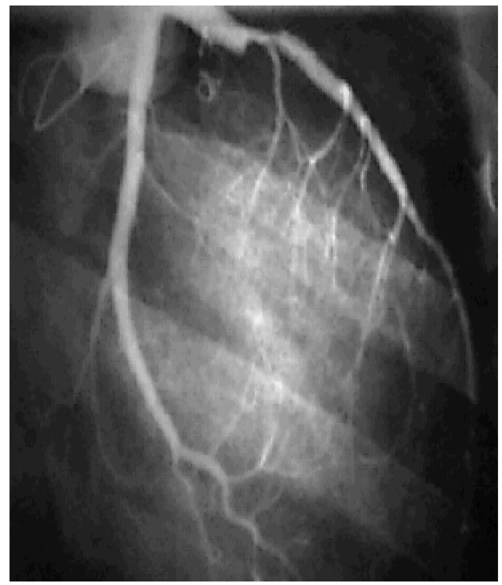
Modalités d' Imagerie

- **X-Ray: Coronarographie**



- Cathétérisme

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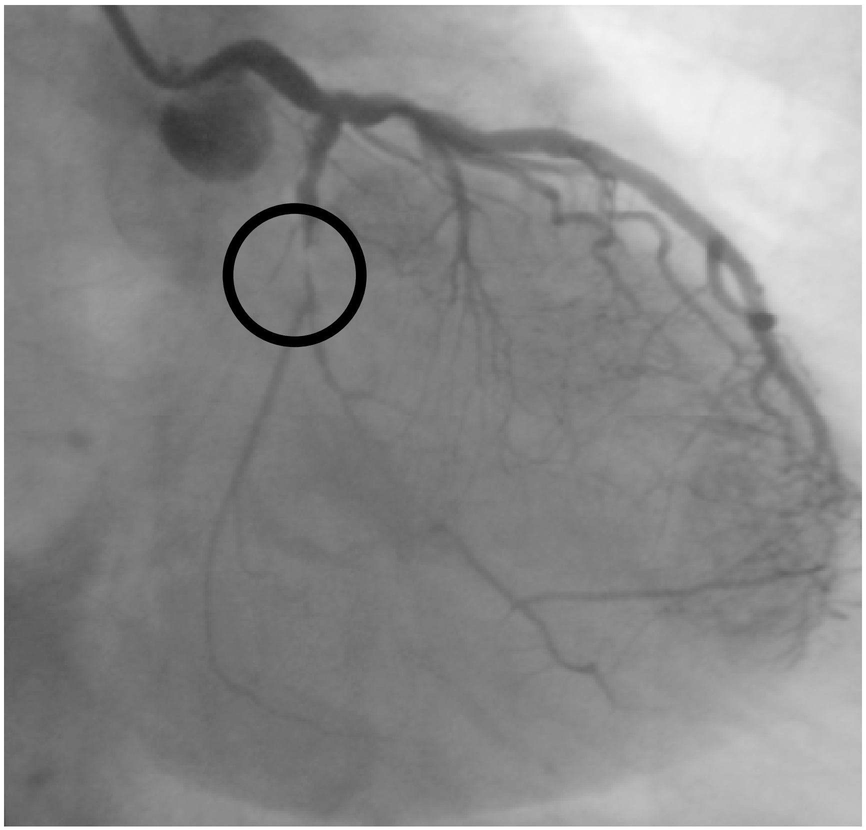
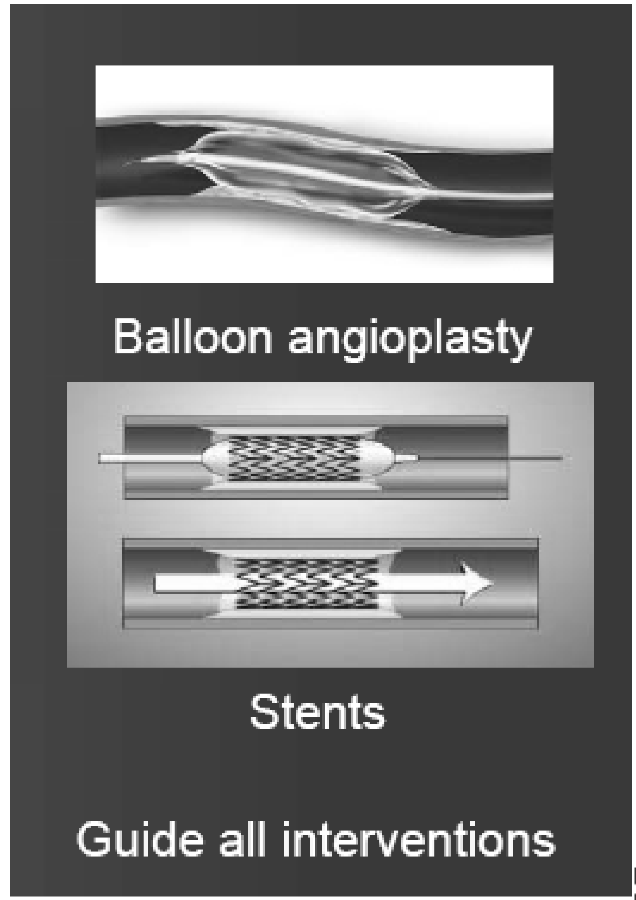
Modalités d' Imagerie

- **X-Ray: Coronarographie**



Imagerie Standard de diagnostic



Imagerie Interventionnelle

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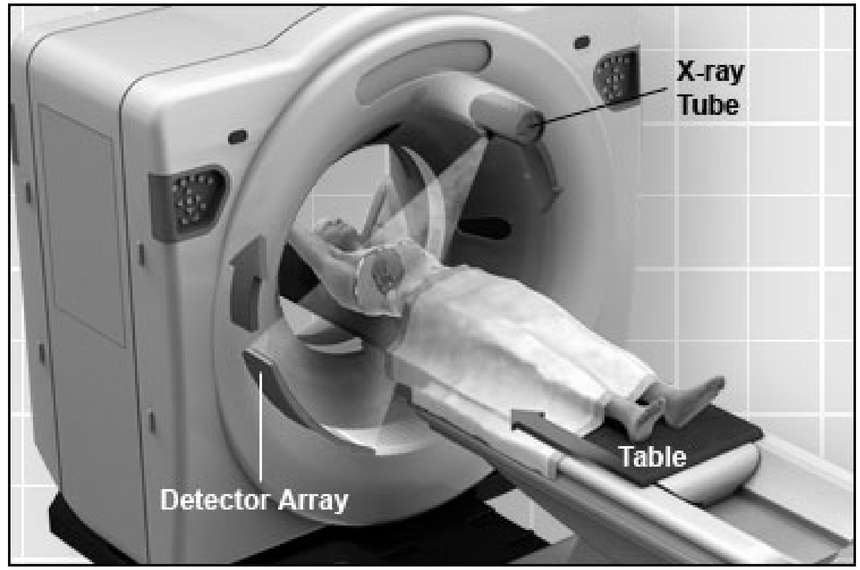
Modalités d' Imagerie

CT Angiographie

- **Résolution spatiale** = 0.4 mm (angiographie coronaire conventionnelle = 0.15-0.25 mm)
- **Résolution temporelle** = 166 msec par rotation (angiographie conventionnelle = 6 msec)
- **64 tranches** en 1 rotation (tranches entrelacées).
- Volume cœur entier en 5-15 secondes.

⇒ Limiter les radiations et les doses d'agent de contraste



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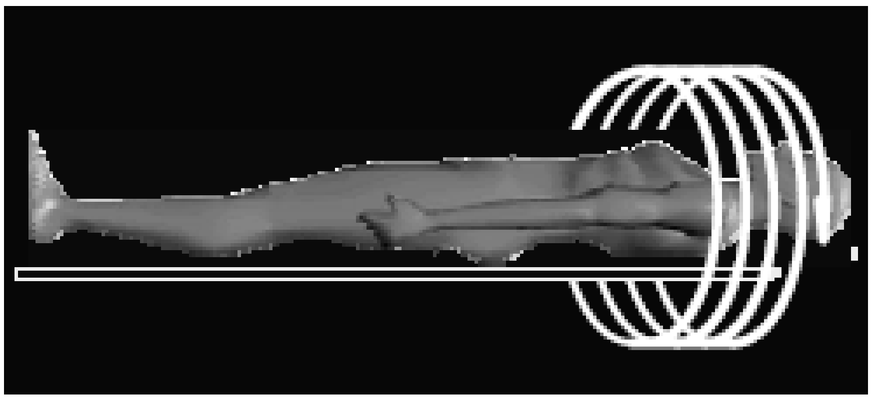
Modalités d' Imagerie

CT Angiographie



Fast rotation speed (3-4 rev/sec)
Multislice acquisition (64 slices/rev)
<1 mm spatial resolution
Fast table translation (175 mm/sec)
Fast acquisition (50 msec/slice)
Cardiac gating capability
Short breath hold for sick patients

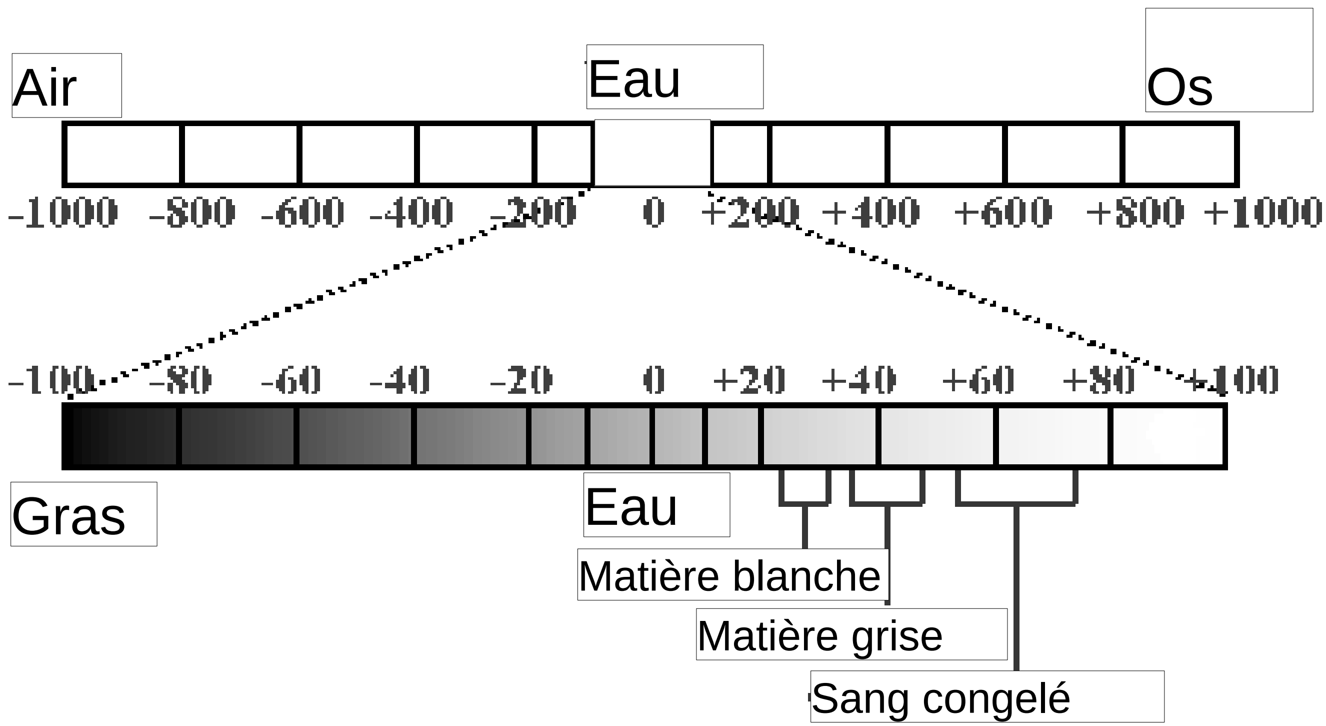
Modalités d' Imagerie

CT

$$HU = 1000 \bullet \frac{\mu - \mu_W}{\mu_W}$$

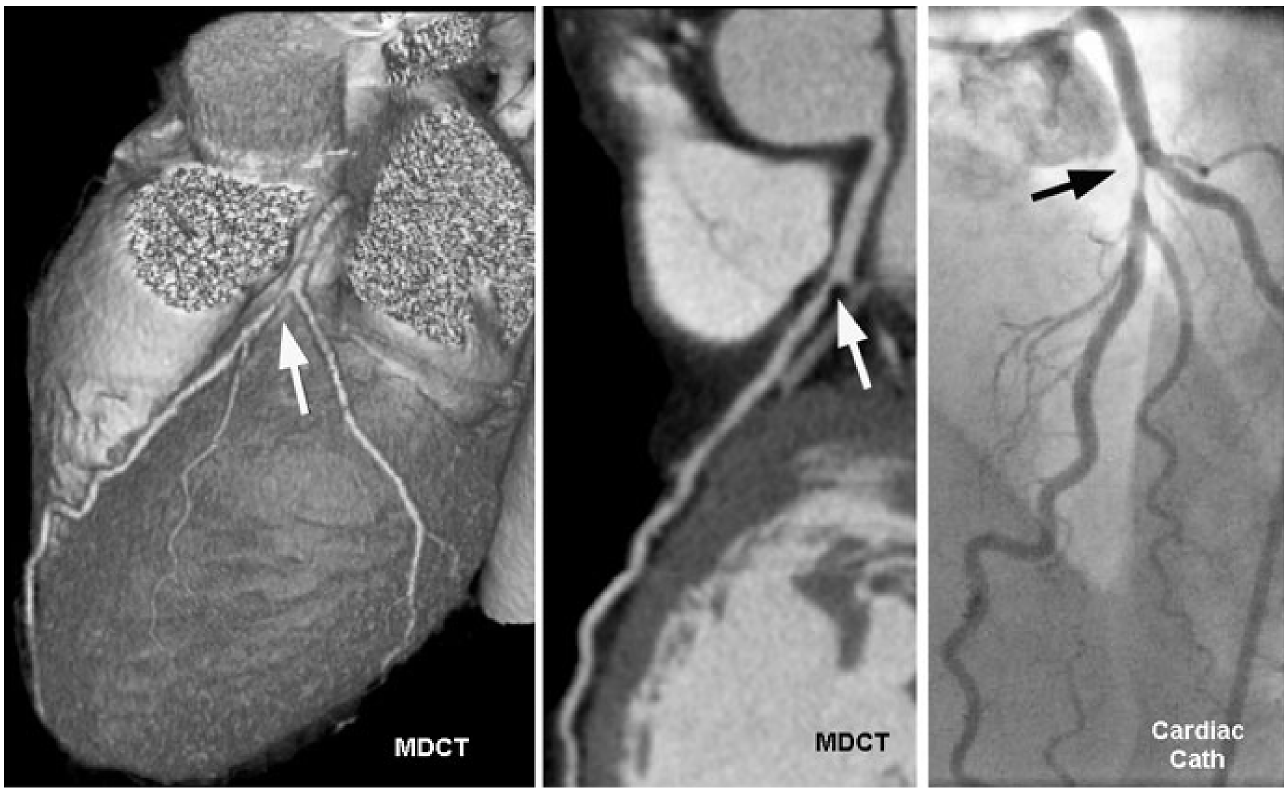
Unité de Hounsfield: Rapport entre le coefficient d'atténuation linéaire mesuré et le coefficient pour l'eau à 50keV.

Valeurs des organes du corps humain.



Modalités d' Imagerie

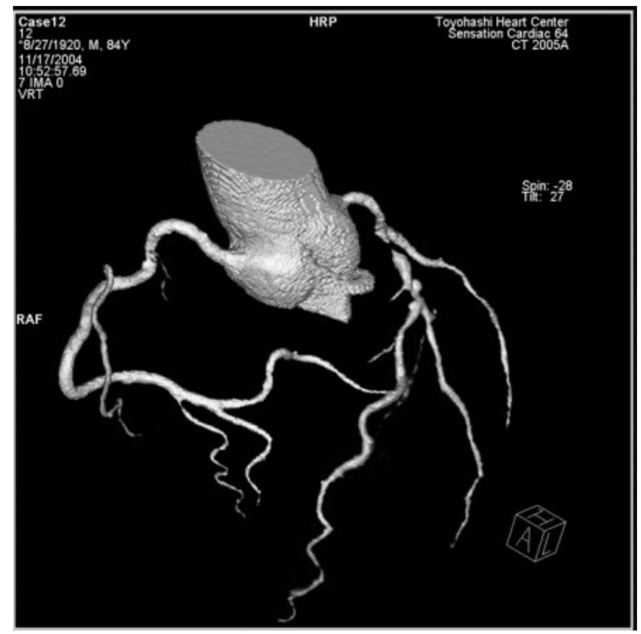
CT Angiographie



Modalités d' Imagerie

CT: Visualisation 3D

« Plan » 3D



Coronaires

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Modalités d' Imagerie

- CT

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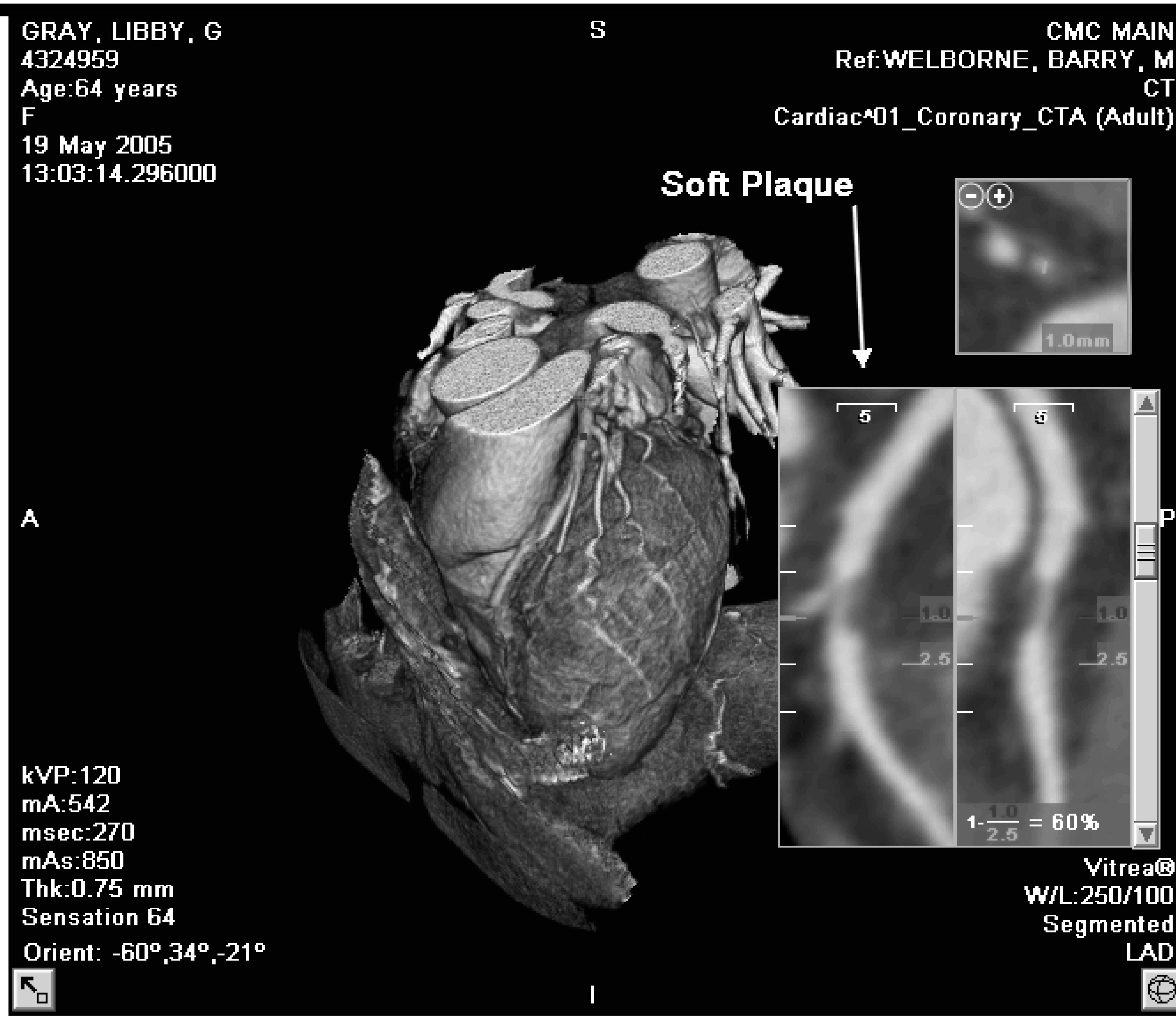
Modalités d' Imagerie



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Modalités d' Imagerie

CT: Conclusions Actuelles

- ***ne remplacera pas*** les examens d'angiographie classiques.
- Post-traitement des images très difficile.
- Très bonne prédiction négative.
- IRM à préférer à cause des radiations

Modalités d' Imagerie

CT: Inconvénients:

- Rythmes cardiaques rapides (>80 bpm) et irréguliers.
- Stents.
- Besoin d'agent de contraste (Cr > 2.0 mg/dl).
- Limitation de la résolution spatiale (<1.5 mm).
- Patients obèses.
- EXPOSITION AUX RADIATIONS.

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Modalités d' Imagerie

CT: Radiations

- PA/Lateral CXR 0.04-0.06 mSv
- Head CT 1-2 mSv
- Chest CT 5-7 mSv
- Abd/Pelvis CT 8-11 mSv
- Diagnostic Coronary Angiogram 3-5 mSv
- MSCT angiography 9.3-11.3 mSv

***Average annual background radiation in U.S ~ 3.6 mSv**

Morin et al. *Circulation* 2003;107:917-22.

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Modalités d' Imagerie

CT: Radiations

- Quantification des effets nocifs des radiations difficile à évaluer.
- **Enfant de moins de 15 ans**: risque de cancer par examen CT = 1 sur 500.
- **Adulte de 45 ans**: risque de cancer par examen CT = 1 sur 1250.

Brenner et al. Radiology, 231(2):440-445.

Modalités d' Imagerie

CT: Prescription

- **Score de calcium**: stratification de risques sur des patients à risques intermédiaires.
- Angiographie des coronaires non-invasive: patients symptomatiques à bas risques ou patients asymptomatiques à risques intermédiaires.

*Un test *négatif (normal CTA) a 98% de chance d'avoir un résultat d'angiographie coronaire normal*

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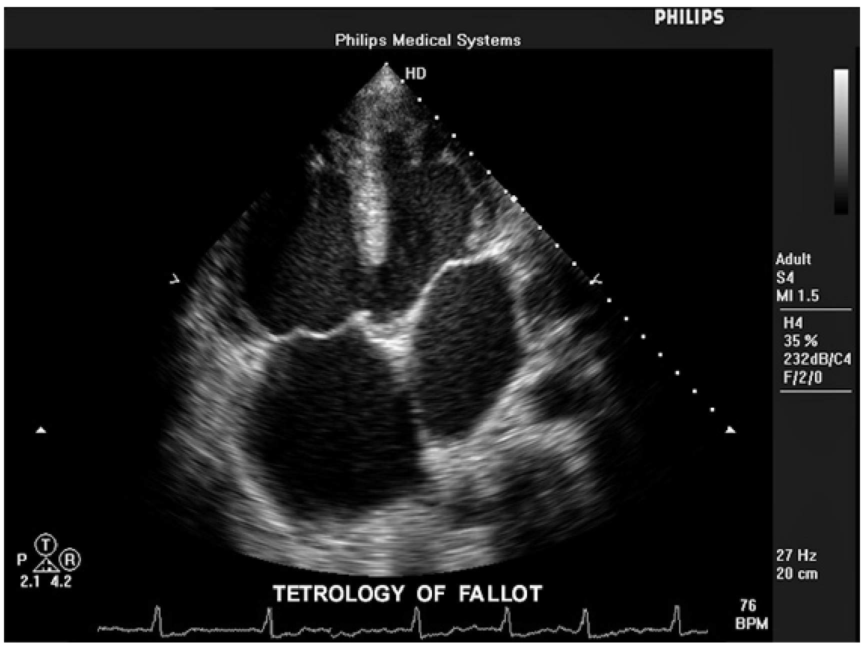
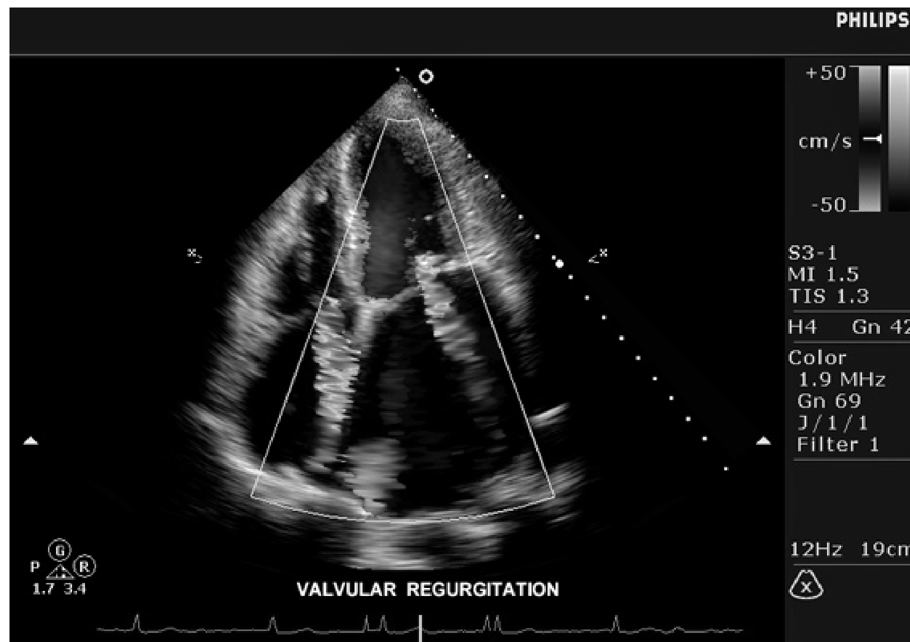
Modalites d' Imagerie

Ultrasons

- Modalité temps-réel, économique, portable, inoffensive.





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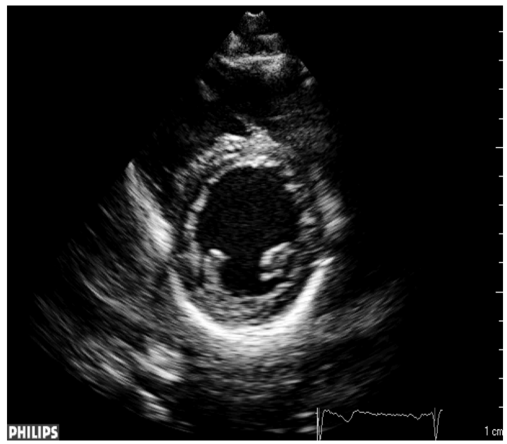
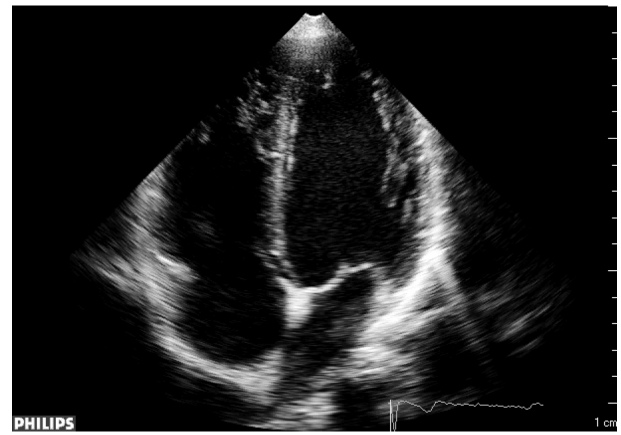
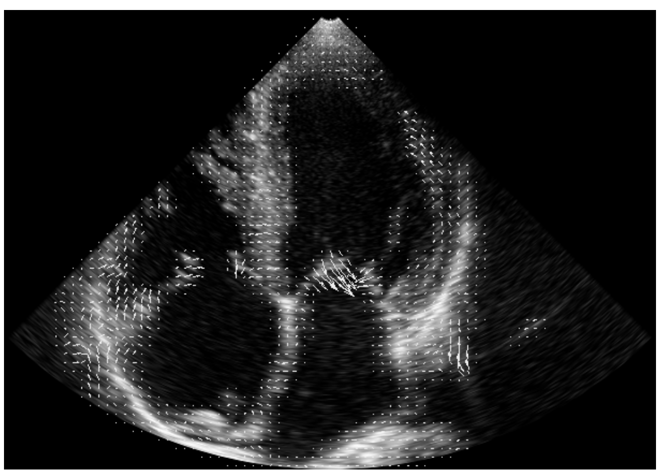
Modalites d' Imagerie

Ultrasons: Temps-réel







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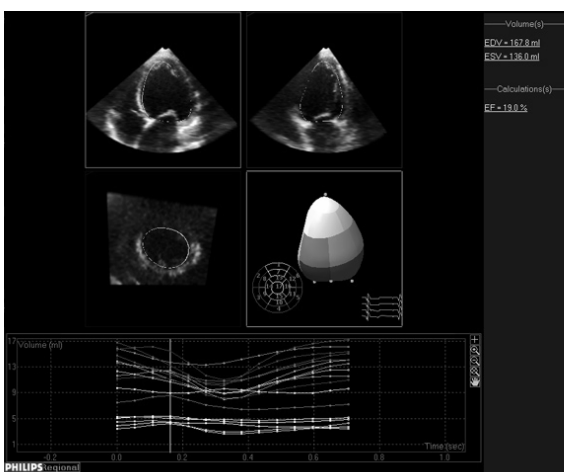
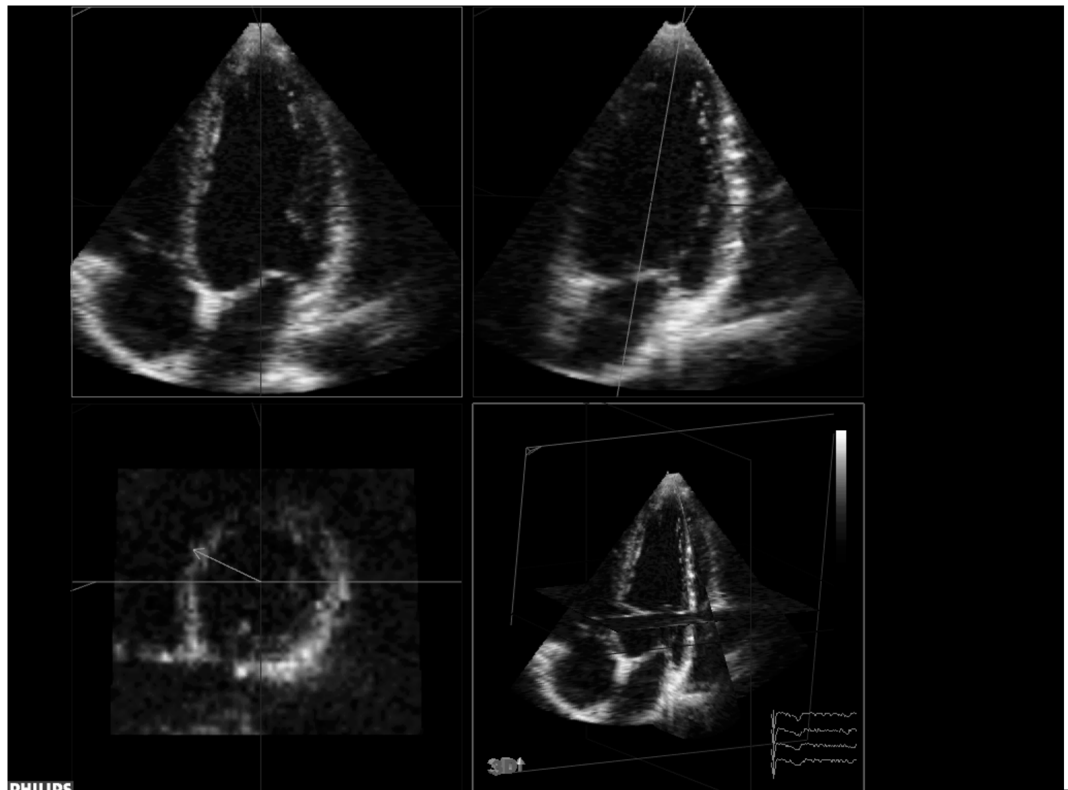
Modalites d' Imagerie

Ultrasons: 3D





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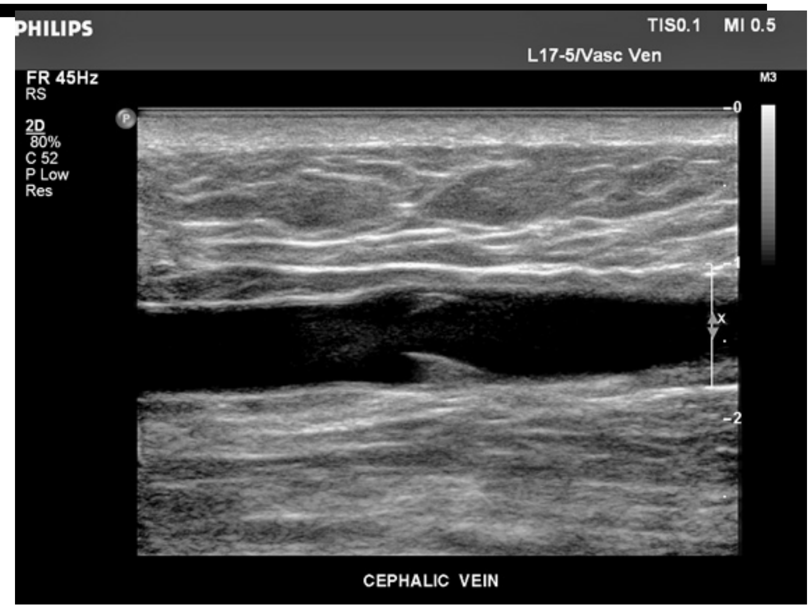
Modalites d' Imagerie

Ultrasons: vasculaire



Modalités d' Imagerie

Ultrasons: Doppler

Exemples d'effet Doppler en médical

- Doppler continu
- Doppler par pulsation: Doppler Spectral

⇨ **Mesures d'indices spectraux**

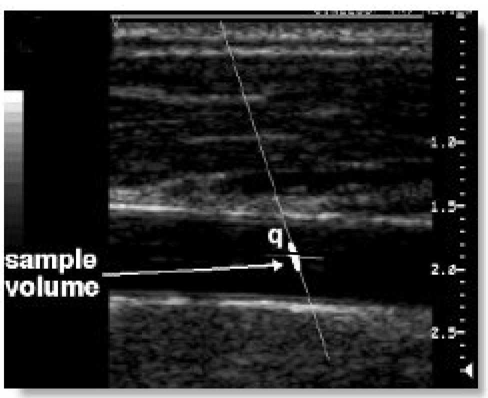


Figure 1

fréquence

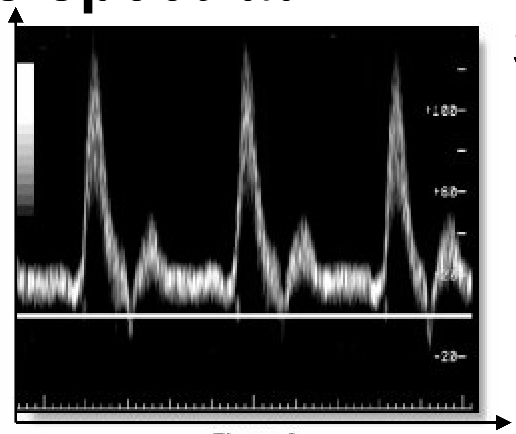


Figure 2

Spectre de Fourier

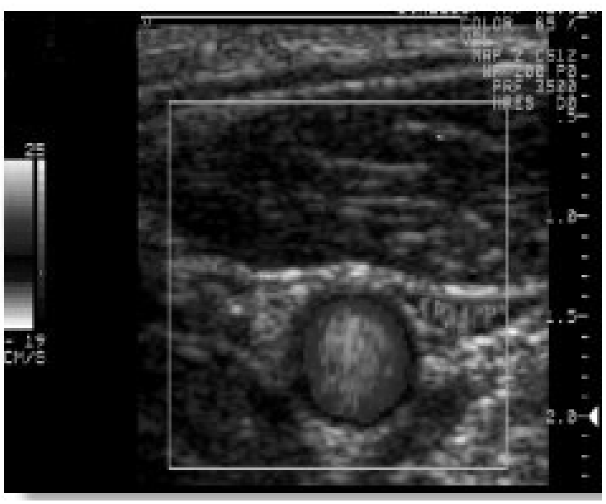
temps

Modalités d' Imagerie

Ultrasons: Doppler

Exemples d'effet Doppler en médical

- Doppler couleur



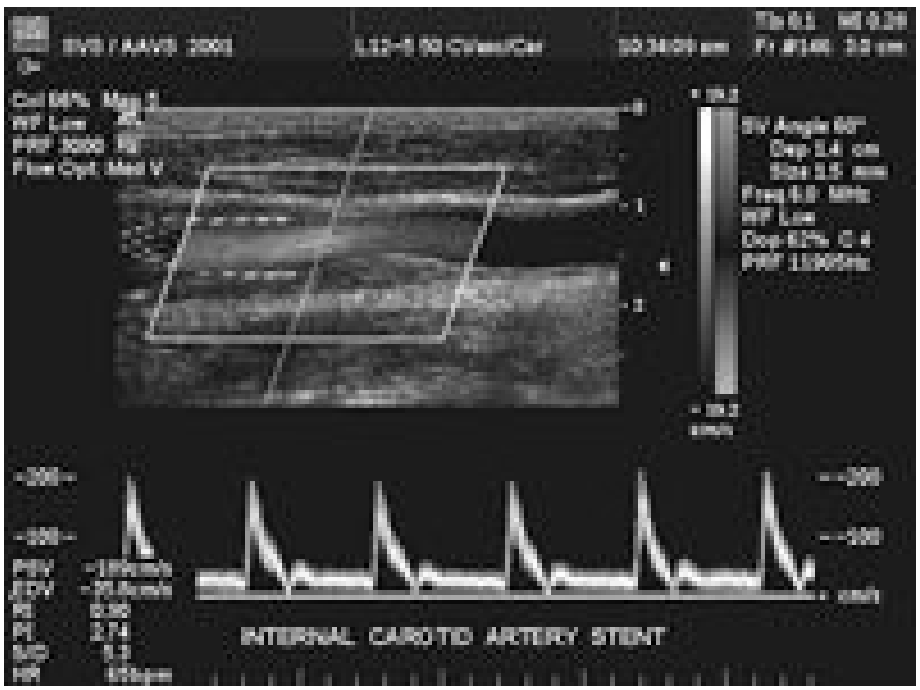
Artère et veine ombilicales

Source: www.gehealthcare.com

Modalités d' Imagerie

Ultrasons: Doppler

- **Exemples de Doppler en Medical**



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Modalités d'Imagerie

Ultrasons: Elastographie / Déformations

- **Elastographie**: mesure de déformation (strain) des tissus par contrainte controllée.
- La réponse des tissus est fonction de leurs propriétés mécaniques.
- **Imagerie de Déformation**: Mesures de déformations (strain) des tissus sous l'effet de déformations physiologiques (pression sanguine, contractions des muscles).
- Différentes méthodes de mesures de déformation:

1. rf ou enveloppe des échos.
2. Tissue Doppler Imaging: contrainte/déformations.
3. Sonoelasticité.

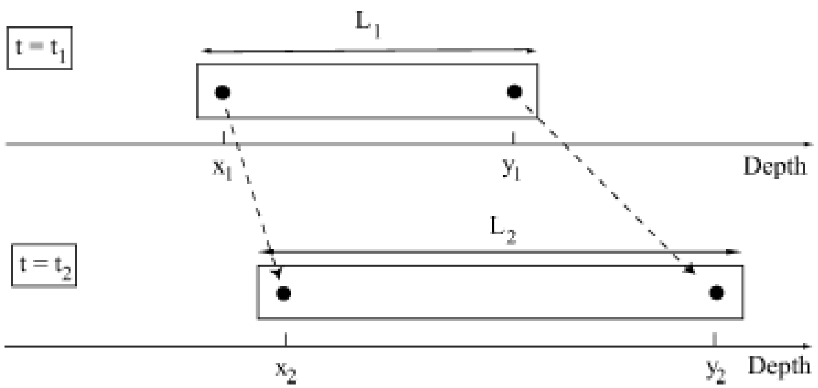
Modalités d'Imagerie

- **Ultrasons: Images de Déformation**

$$\sigma(t) = E \times \varepsilon(t)$$

Force totale (wall stress) — **Elasticité** — **Déformation**

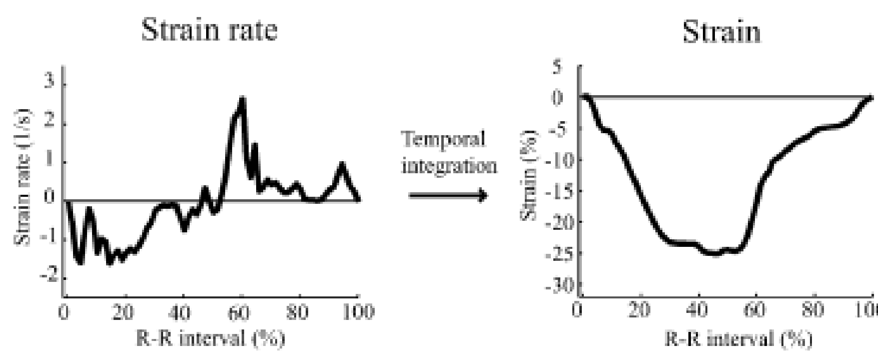
ε(t): Déformation d'un objet par rapport à sa forme de référence.



$$\varepsilon = \frac{L_2 - L_1}{L_1} = \frac{(y_2 - x_2) - (y_1 - x_1)}{y_1 - x_1}$$

Mesures par ultrasons: **taux** de déformation

$$\dot{\varepsilon} \approx \frac{\varepsilon}{\Delta t} = \frac{\frac{(y_2 - y_1)}{\Delta t} - \frac{(x_2 - x_1)}{\Delta t}}{y_1 - x_1} \approx \frac{v_2 - v_1}{L_1}.$$



Modalités d'Imagerie

- **Ultrasons: Images de Déformation**

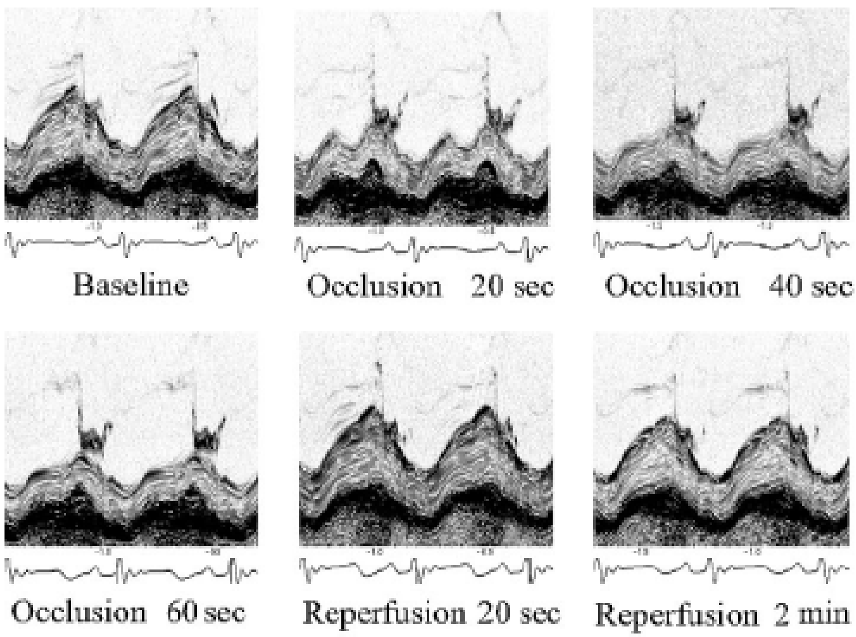


Fig. 6. M-mode images taken from the posterior wall in a parasternal long axis view. The change in wall deformation during acute ischemia and subsequent reperfusion can clearly been appreciated.

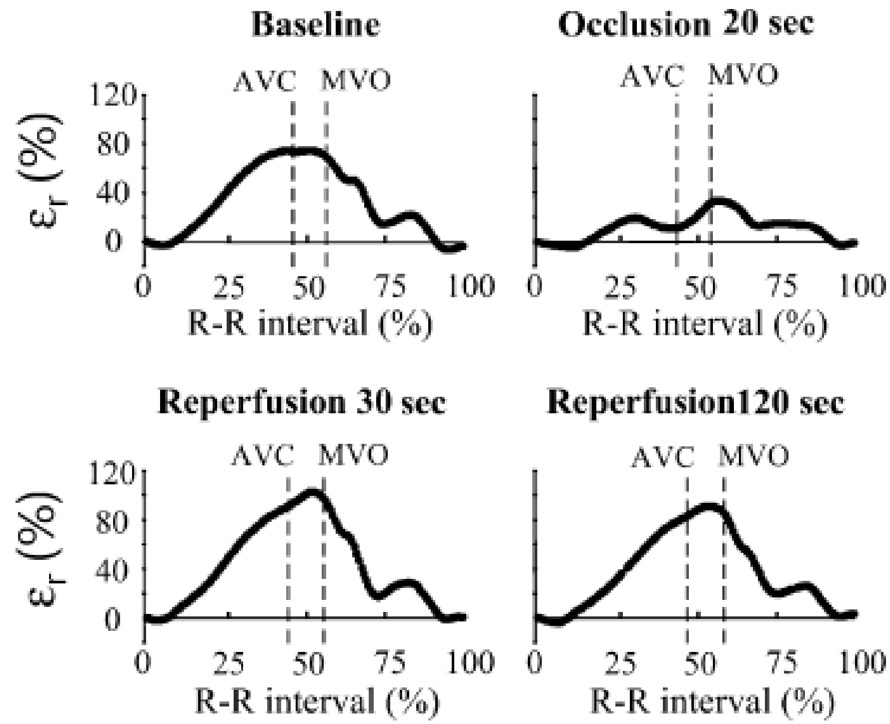


Fig. 7. Strain measurements during experimental acute ischemia. The two vertical lines in the middle of the curves indicate the time point of aortic valve closure (AVC) and mitral valve opening. At baseline, the myocardium is thickening before AVC, whereas during ischemia, deformation is blunted and some thickening occurs after AVC. With reperfusion, after a short period of hypercontraction, strain returns to normal.

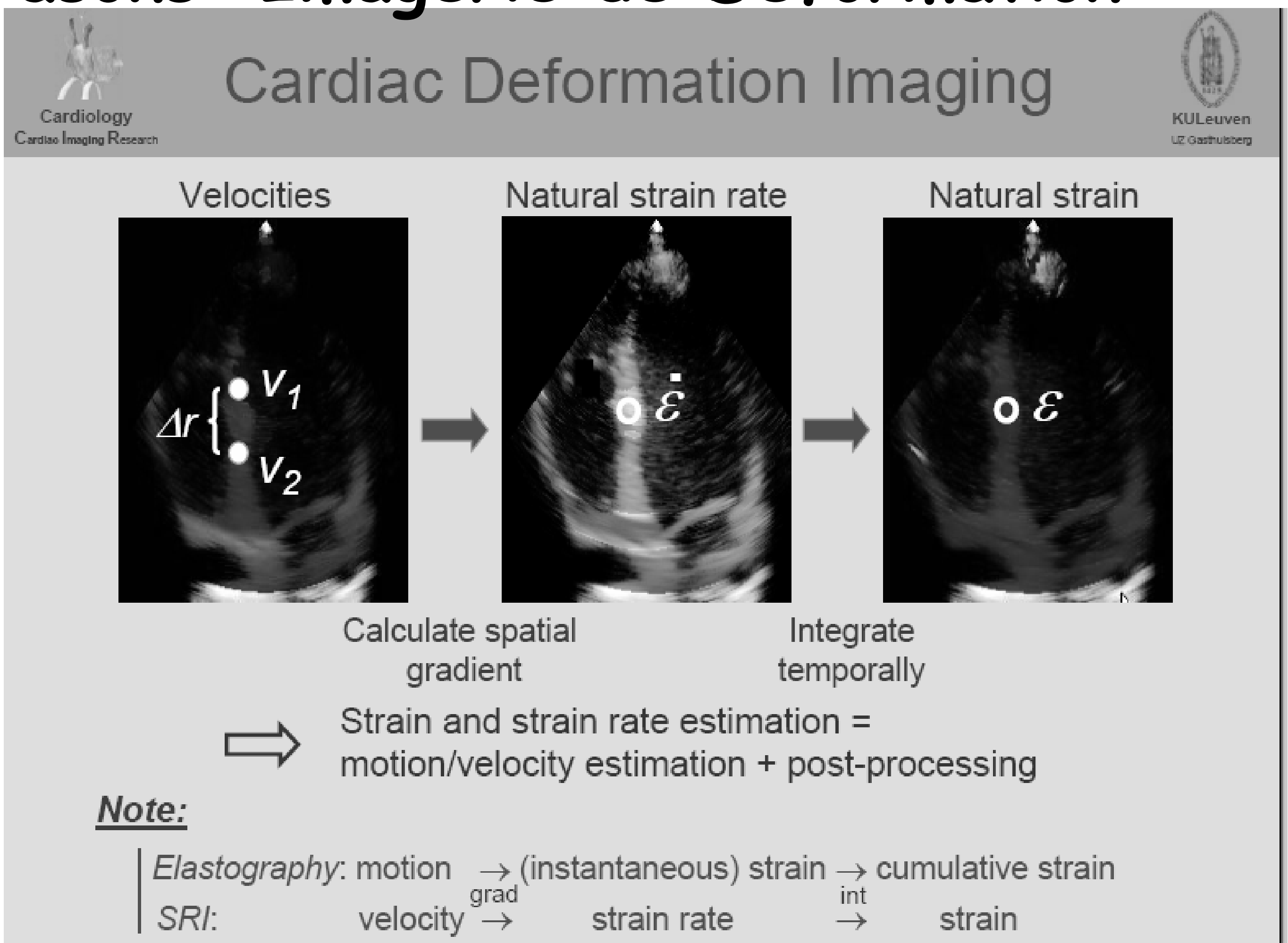
1022 IEEE TRANSACTIONS ON MEDICAL IMAGING, VOL. 21, NO. 9, SEPTEMBER 2002

Echocardiographic Strain and Strain-Rate Imaging: A New Tool to Study Regional Myocardial Function

Jan D'hooge*, Bart Bijnens, Jan Thoen, Frans Van de Werf, George R. Sutherland, and Paul Suetens

Modalités d'Imagerie

- **Ultrasons: Imagerie de Déformation**



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Modalités d'Imagerie

- **Ultrasons: Imagerie de Déformation**

Cardiology
Cardiac Imaging Research

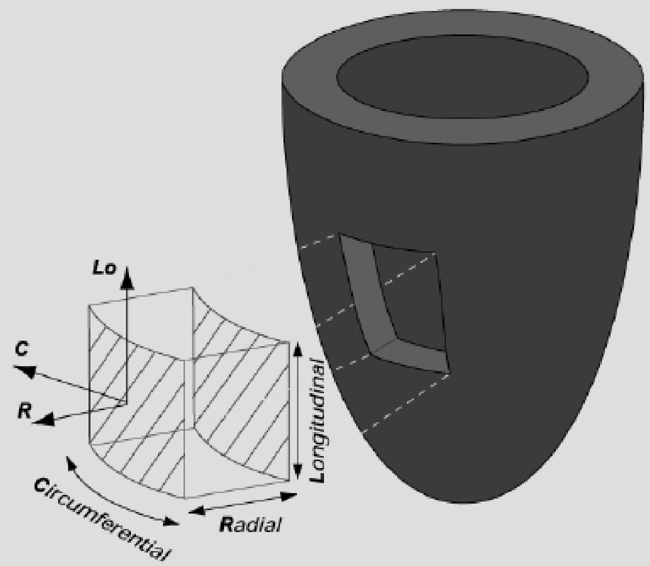
Local cardiac coordinates

KULeuven
UZ Gasthuisberg

- Radial
- Longitudinal
- Circumferential

⇩

$$\varepsilon = \begin{bmatrix} \varepsilon_{RR} & \varepsilon_{RC} & \varepsilon_{RL} \\ \varepsilon_{CR} & \varepsilon_{CC} & \varepsilon_{CL} \\ \varepsilon_{LR} & \varepsilon_{LC} & \varepsilon_{LL} \end{bmatrix}$$



Facilitates physical interpretation and mathematics of the strain values
(e.g. RR = wall thickening; CC/LL = circumferential/longitudinal shortening)

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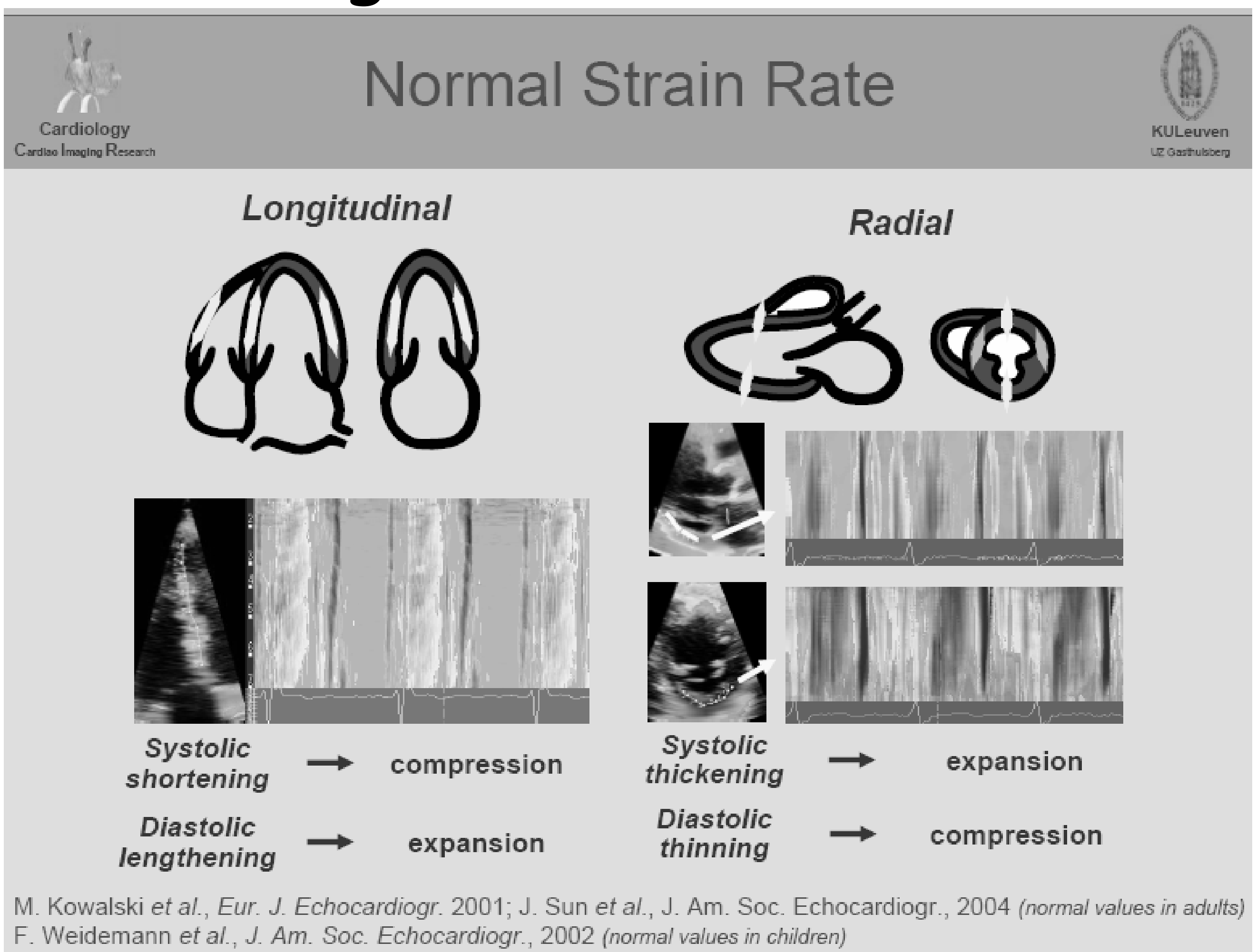
Modalités d'Imagerie

- **Ultrasons: Imagerie de Déformation**



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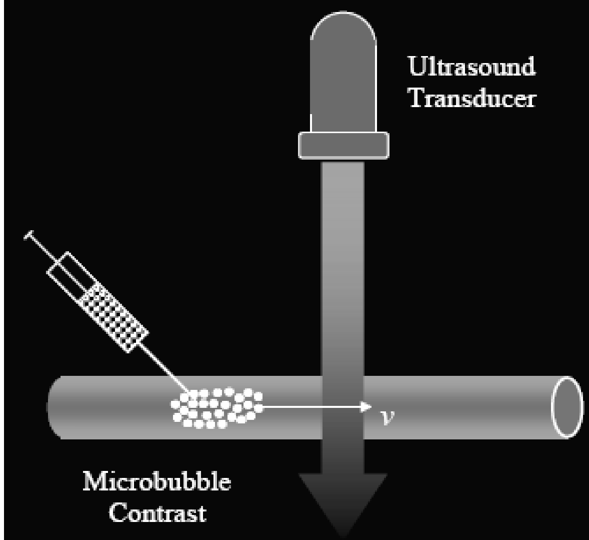
Modalités d'Imagerie

- Ultrasons: Agent de Contraste

Myocardial Contrast Echocardiography (MCE)

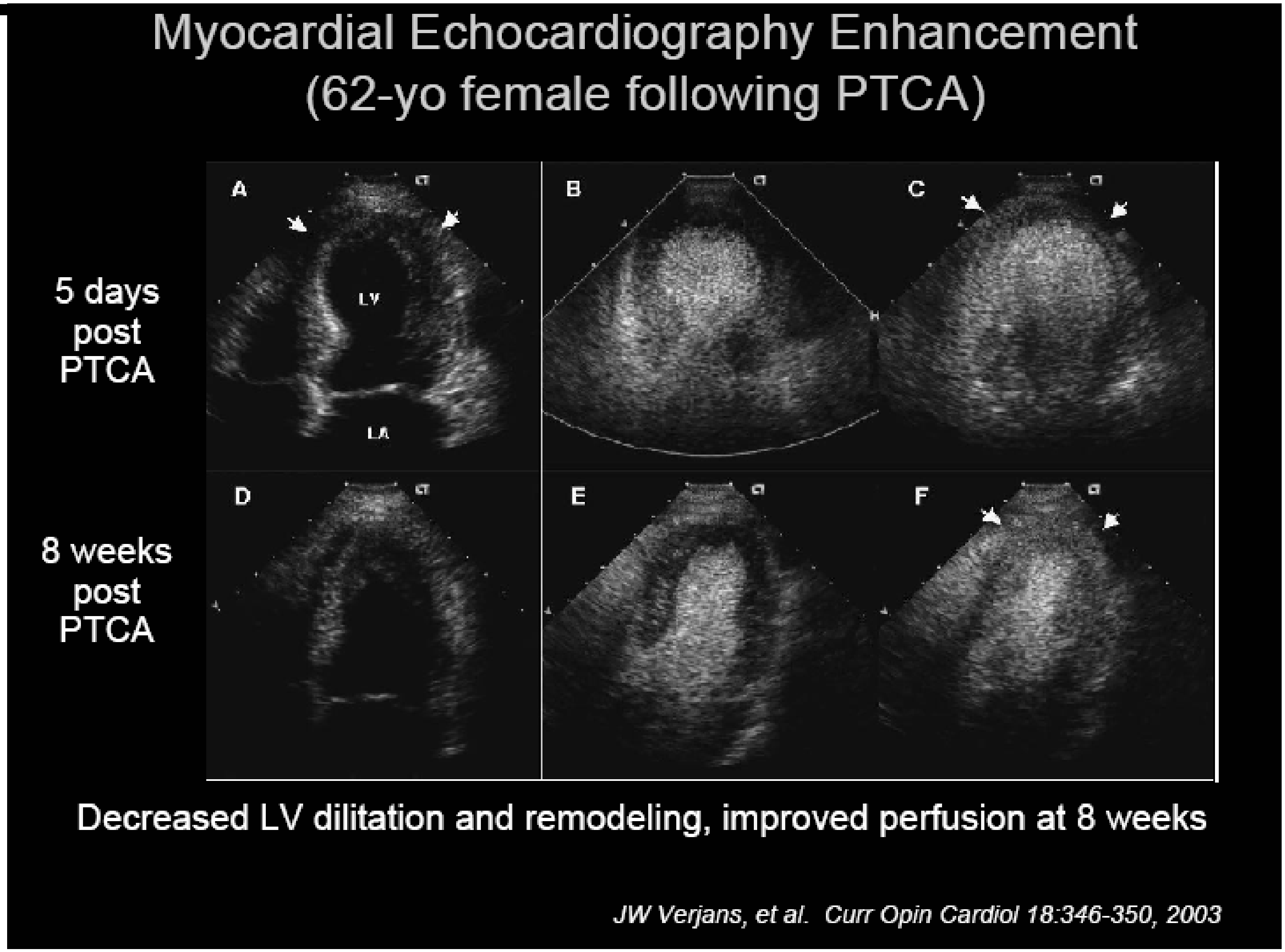


MICROBUBBLES

- Encapsulated gas bubbles
- 1-7 microns in diameter
- Can change diameter 2- to 10-fold
- Injected IV then pass through pulmonary vasculature into arterial circulation
- Natural echogenicity with resonant frequency of 2-10 MHz

Modalités d'Imagerie

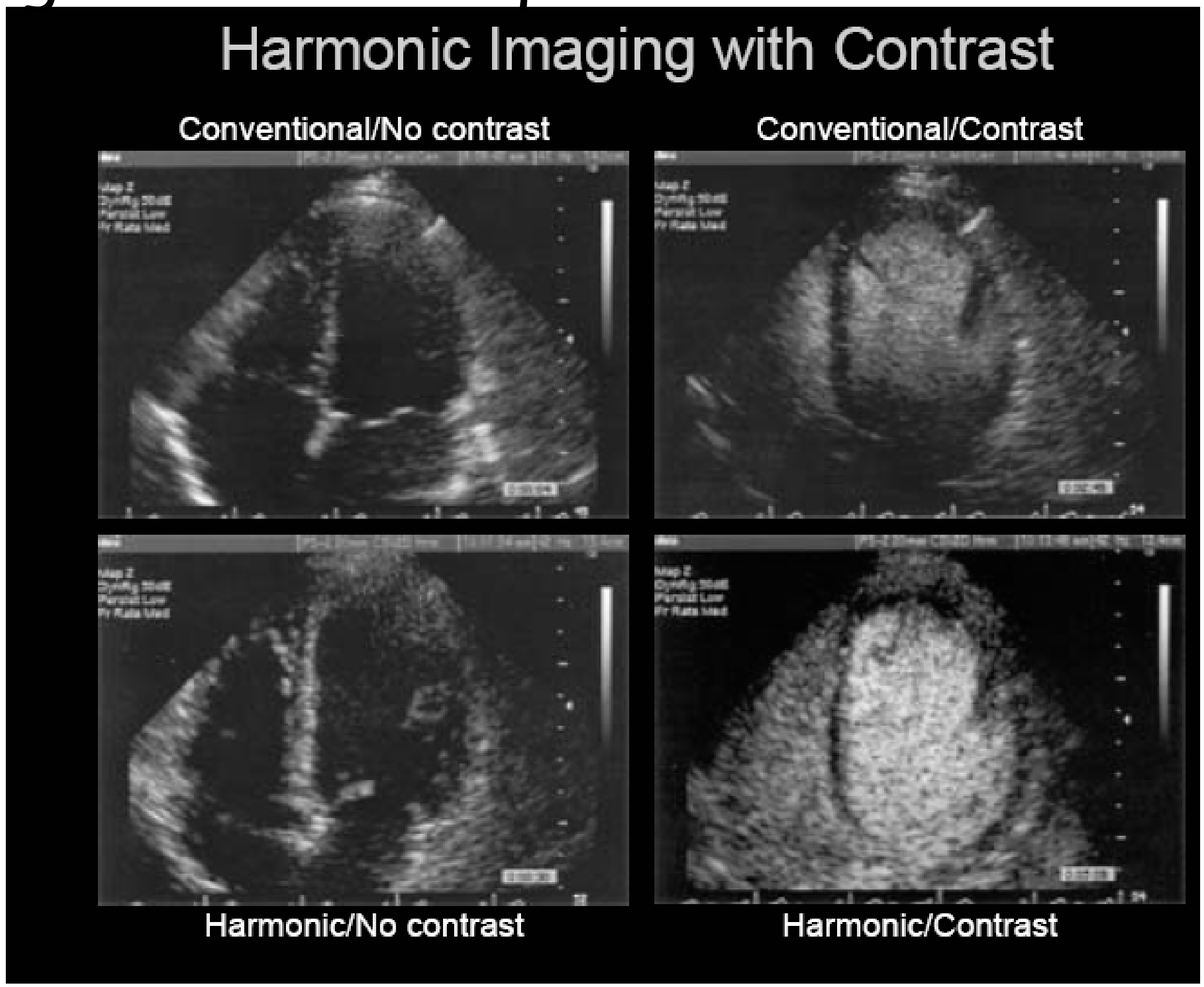
- Ultrasons: Agent de Contraste



PTCA = Percutaneous Transluminal coronary angioplasty

Modalités d'Imagerie

- Ultrasons: Imagerie Harmonique



Physics of Cardiovascular imaging, B. Hasegawa, UCSF

Modalités d'Imagerie

- IRM
  - Nombreux protocoles d'acquisition (CINE).
  - Antenne de surface.
  - Apnée.
  - Sang noir ou blanc.
  - Résolution ~$1.5mm^2$*8mm.
  - Synchronisation par ECG.
  - ~16 images par cycle.
  - Agent de contraste (gadolinium) + rehaussement tardif.

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Modalités d'Imagerie

- IRM

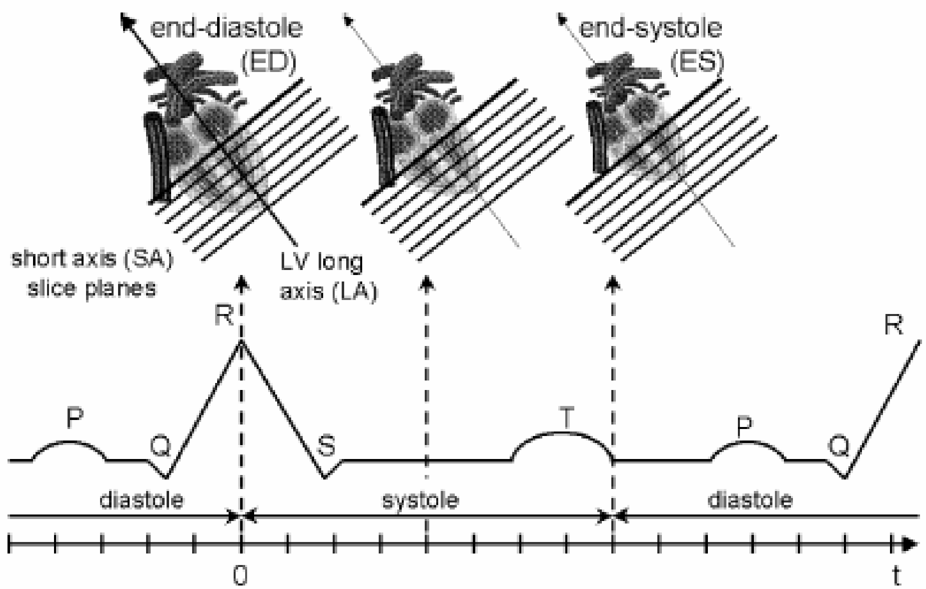

Fig. 1. Illustration of a classical acquisition of SA images with a ECG-gated cine MR sequence. The same slice is acquired at successive time points of the cardiac cycle. However, due to the motion of the heart, we do not observe the same anatomical region within the same slice. Moreover, several cardiac cycles are required to reconstruct slices. When possible, patients are asked to retain their breath (15–20 s) during acquisition.

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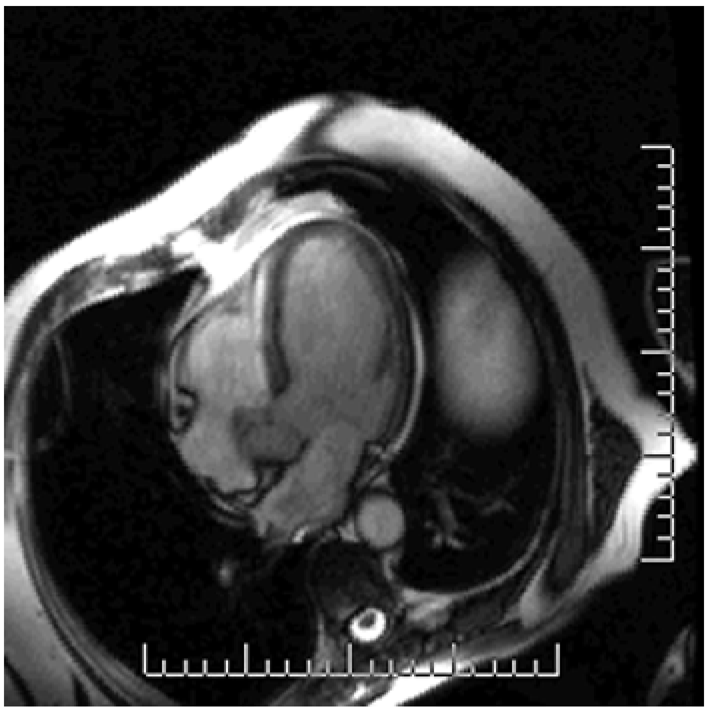
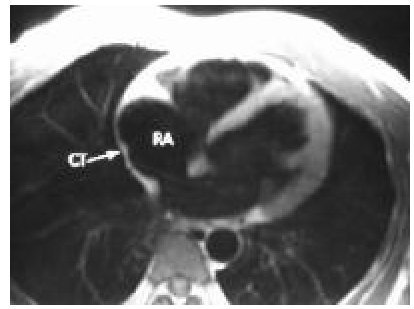
Modalités d'Imagerie

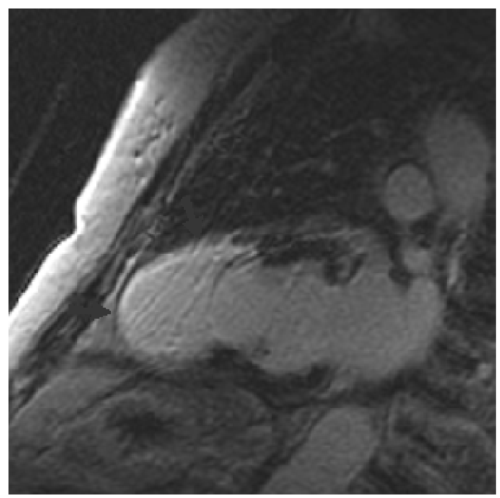
- IRM

Sang noir
ou blanc



Courtesy of Robert R. Edelman





Artéfacts de mouvement:

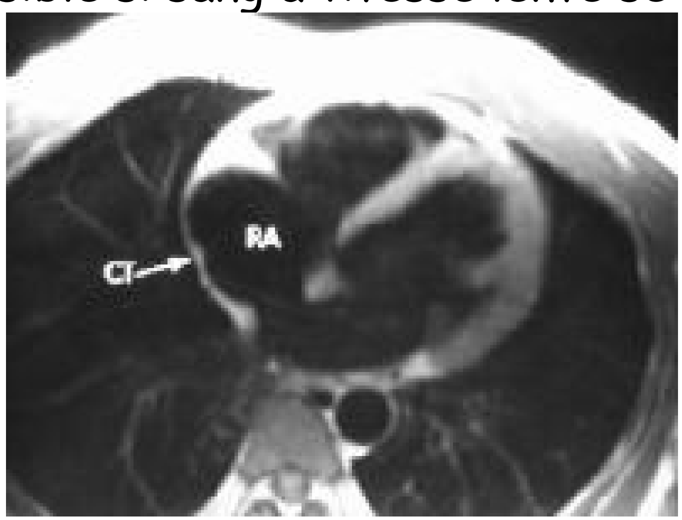
-Synchronisation
cardiaque

-Synchronisation
respiratoire

Modalités d'Imagerie

- IRM: Morphologique

-Sang noir = suppression du signal du sang
- Motif de préparation de l'aimantation par double inversion :
  - 1ere impulsion d'inversion de 180° sur l'ensemble du volume exploré.
  - 2eme impulsion d'inversion de 180° sélective de la coupe à imager.
- Acquisition lancée après un temps d'inversion TI correspondant à l'annulation du signal du sang : le signal du sang entrant dans la coupe est supprimé.
-Signal sanguin possible si sang à vitesse lente se déplaçant dans le plan de la coupe.



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Modalités d'Imagerie

- IRM: Ciné

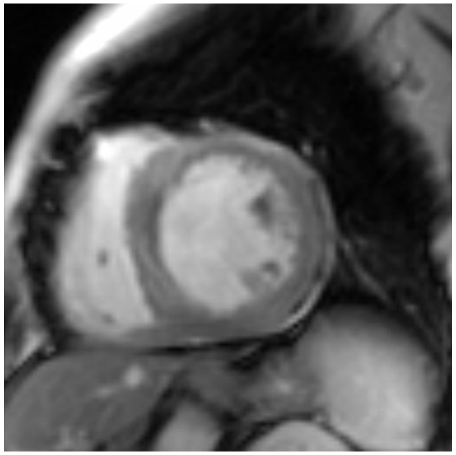
-Séquences d'écho de gradient ultra-rapides, SNR élevé,

- Contraste T2/T1 permettant de bien différencier:

    - sang (en hypersignal),

    - endocarde & épicarde (en isosignal)

    - graisse (en hypersignal).



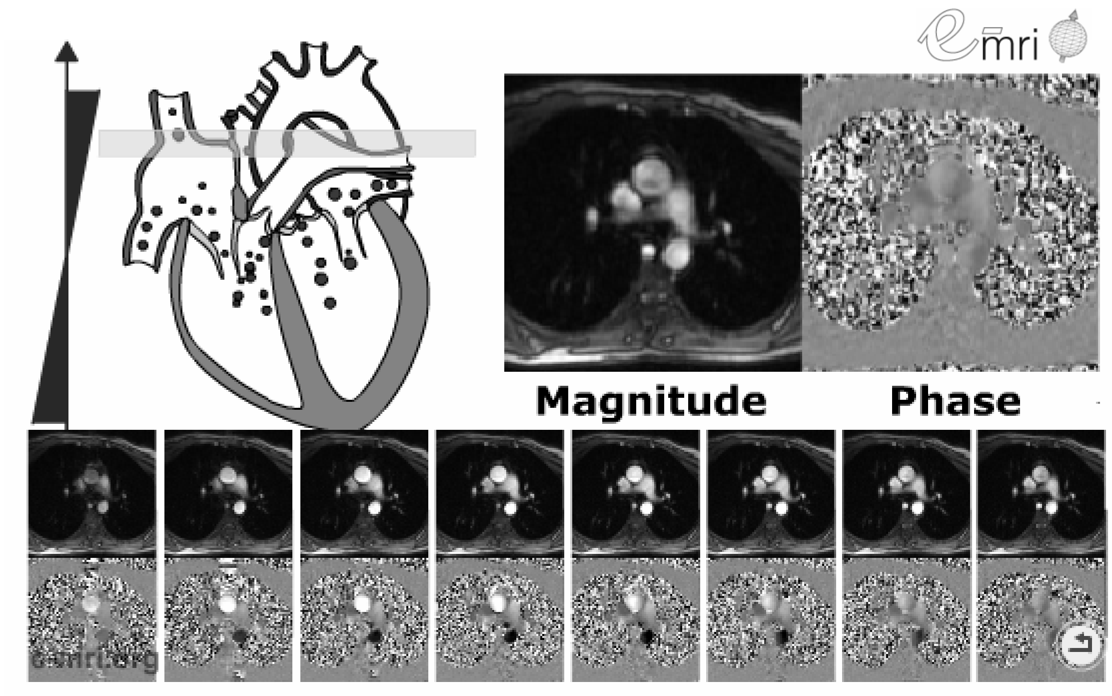
- Gating rétrospectif possible pour amélioration de la résolution temporelle.

Modalités d'Imagerie

- IRM: Imagerie de Flux par Phase

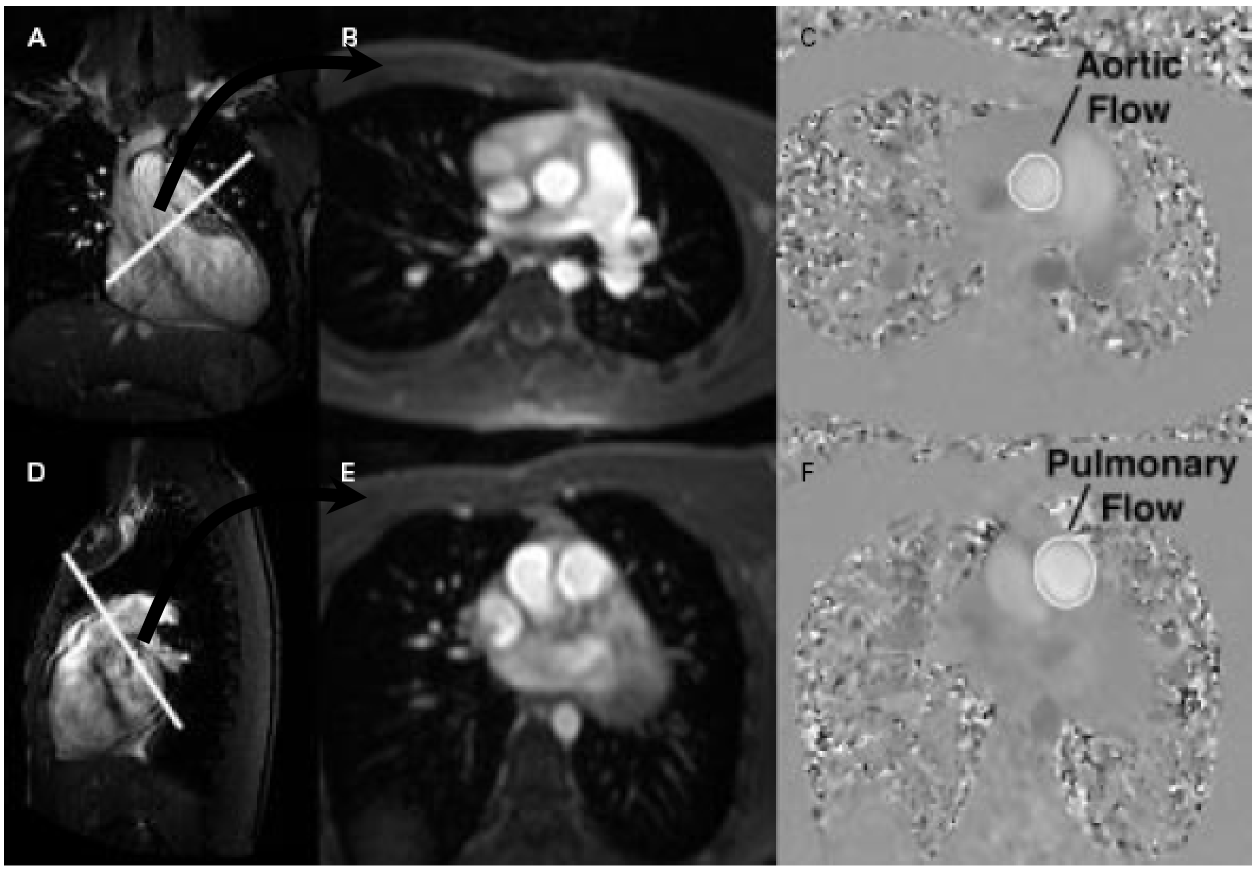
- Imagerie des flux en contraste de phase.
  ⇒ Quantification des vitesses et des débits sanguins.

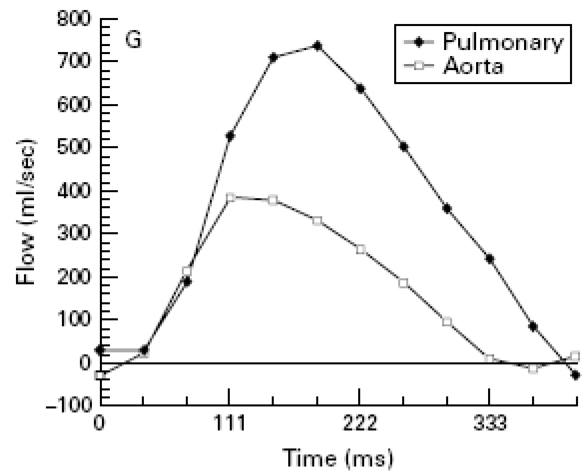
- Acquisition en contraste de phase =
  - images en magnitude "anatomique "
  - images en phase "quantitatives " : direction et vitesse des flux.
- Choix préalable d'une vitesse d'encodage adaptée impératif (aliasing).



Modalités d'Imagerie

- IRM: Imagerie de Flux par Phase

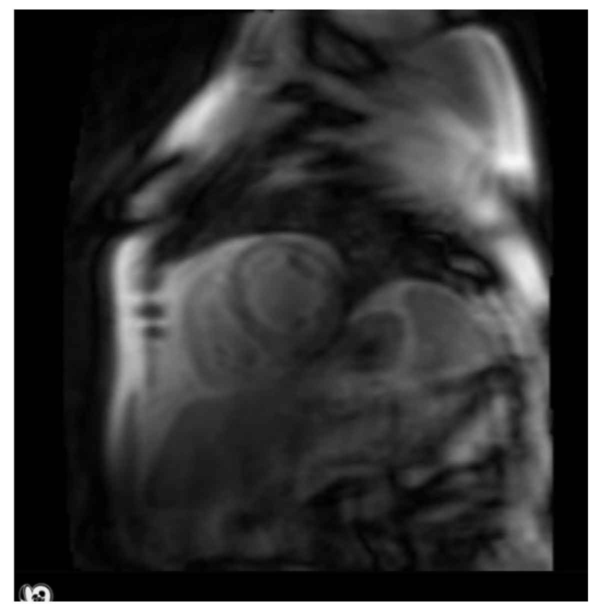




igure 16 Pulmonary and systemic arterial flow letermination by phase contrast magnetic resonance maging in a 17 year old patient with partial anomalous ulmonary venous connection. (A) Coronal positioning can for aortic acquisition, showing position of aortic phase ontrast section. (B) Magnitude image from aortic phase ontrast section. (C) Corresponding phase image from ortic section. (D) Sagittal positioning scan for pulmonary cquisition, showing position of pulmonary phase contrast ection. (E) Magnitude image from pulmonary phase ontrast section. (F) Corresponding phase image from pulmonary section. (G) Calculated systemic and pulmonary flow rates. Calculated Qp:Qs ratio was 2.17:1.

Modalités d'Imagerie

- IRM de rehaussement tardif
  - **Principe:** Etude de la viabilité du myocarde supposant que les zones infarctées sont réhaussées 10-15 minutes après l'injection d'un agent de contraste paramagnétique.
    - Réhaussement par accumulation de gadolinium dans l'espace extracellulaire à cause de la perméabilité de la membrane.
  - **Mesures:** Quantification de la taille, localisation et étendue transmurale de l'infarctus.
    - Si < 50% de l'épaisseur du muscle: contractilité améliorée sur le segment après revascularisation.



Modalités d'Imagerie

- IRM marquée (tagging)
  - **Principe:** Production de bandes de **saturation** (tag) par pulsations de radiofrequences.
  - Imagerie multiphasée pour détecter des diminutions de saturation
  - **Mesures**:
    - Suivi du mouvement des lignes de taggues pendant le cycle cardiaque.
    - Quantification de déplacements de points dans le myocarde.
    - Calcul de déformation.
  - **Problème**: Détection de lignes de taggues et des intersections.

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Modalités d'Imagerie

- IRM marquée (tagging)

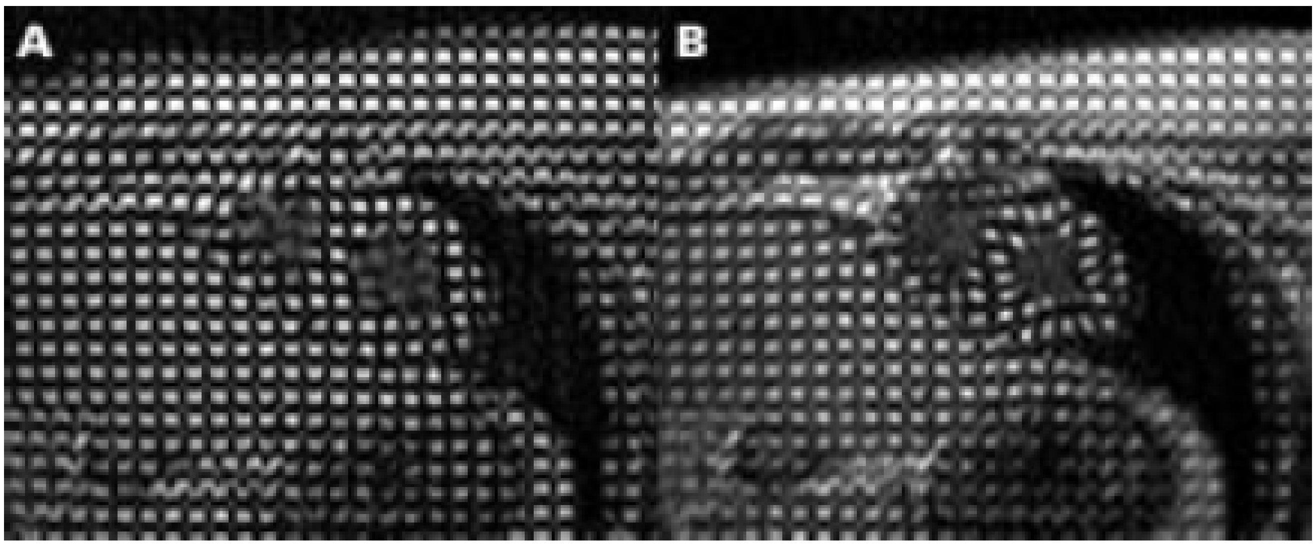
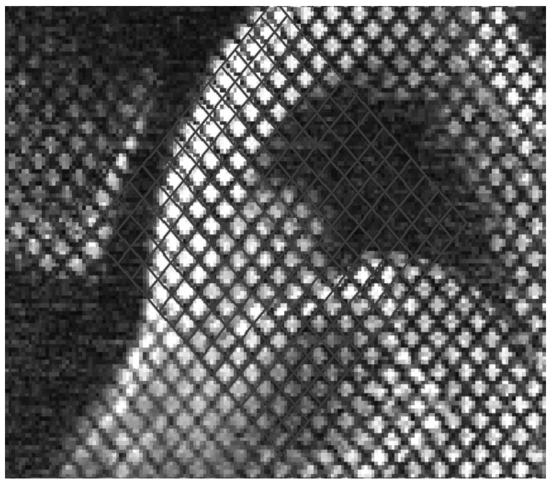
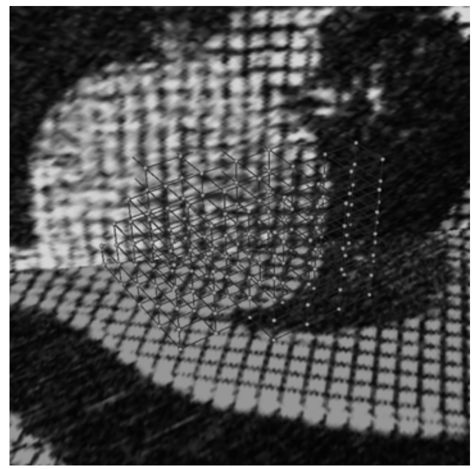


Figure 17 Tagged short axis images at the mid ventricular level in 16 year old patient with normal heart. (A) Diastolic image. (B) Systolic image. Note the distortion of the tagging lines on the systolic frame.

Modalités d'Imagerie

- IRM marquée (tagging)





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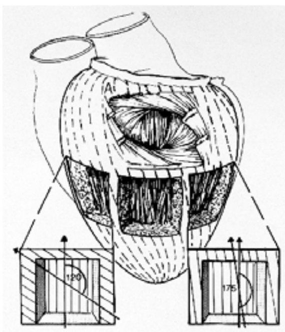
Modalités d'Imagerie

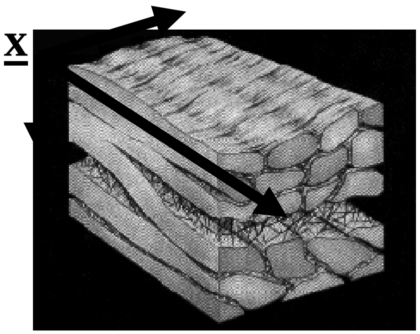
- IRM de Diffusion (DTI)

Diffusion Tensor MR Imaging (DTMRI)



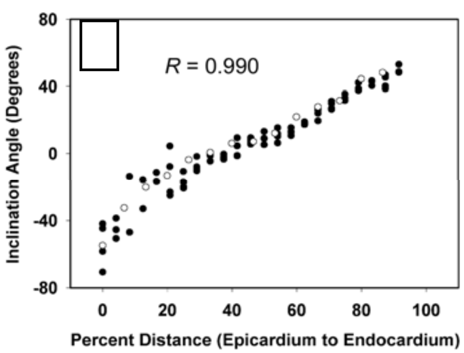
Fox and Hutchins (1972). Johns Hopkins Med. J. 130(5): 289-299

x



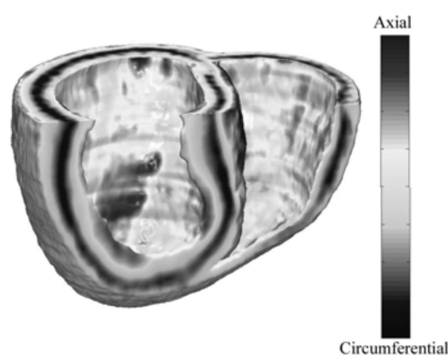
- **DTMRI ⇒ 3x3 diffusion tensor $M_i(\underline{x})$**
- **Hypothesis – The principle eigenvector of $M_i(\underline{x})$ is aligned with fiber direction at point $\underline{x}$**

DTMRI vs HISTO Fiber Angles



Scollan et al (1998). *Am. J. Physiol.* 275: H2308
Holmes, A. et al (2000). *Magn. Res. Med.*, 44:157

DTMRI Reconstruction of Canine Ventricles



Scollan et al (2000). *Ann. BME*. 28(8): 934
Helm et al (2004). *Mag. Res. Imaging*, in review
Beg et al (2004). *Mag. Res. Med.*, in review

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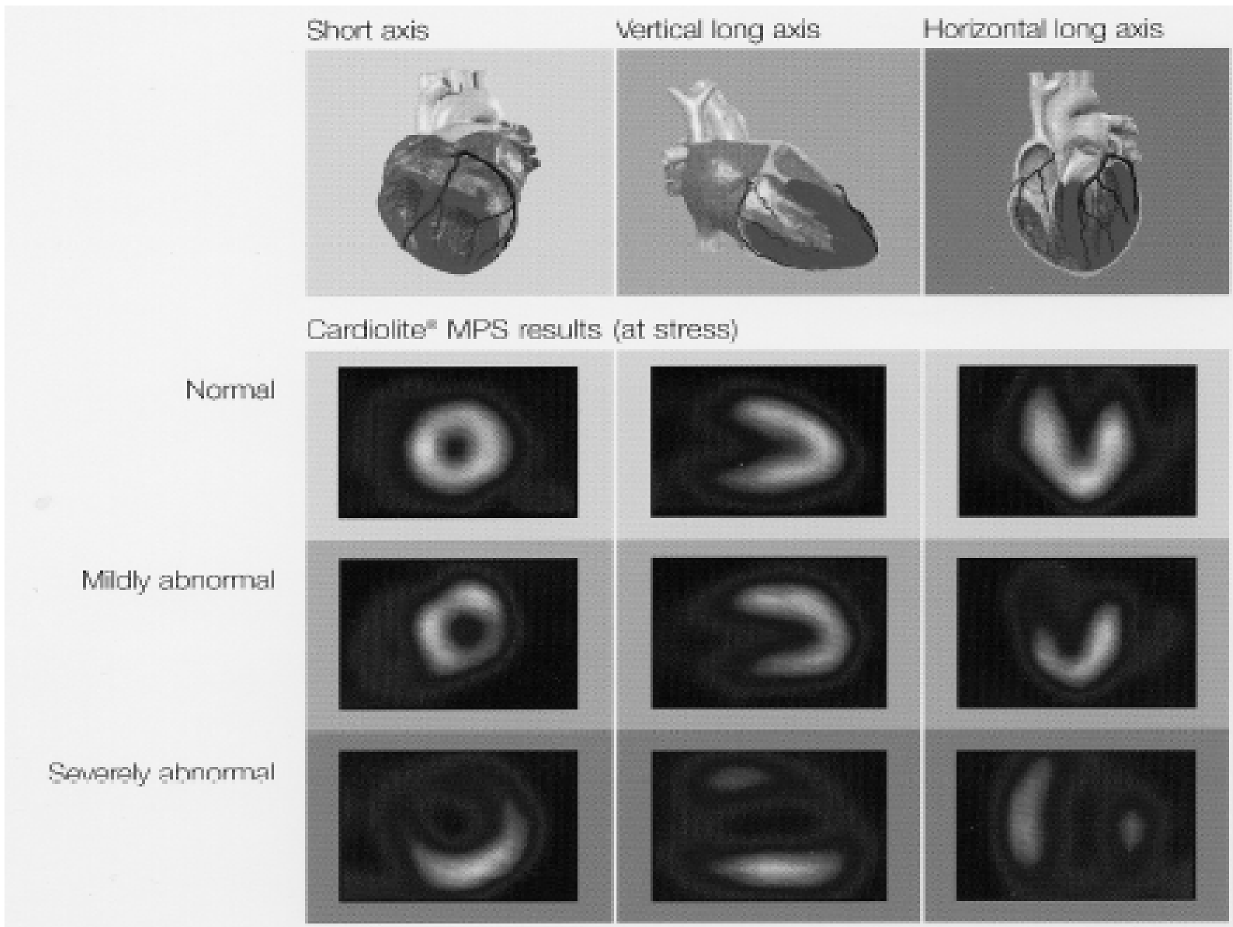
Modalités d'Imagerie

- Nucléaire
  - Imagerie de perfusion myocardique.
  - Utilisation d'agents radioactifs (sur globules rouges)
    - Thallium – 201
    - Technetium-99 m Sestamibi
    - Technetium-99 m Tetrofosmin

Modalités d'Imagerie

- Nucléaire

Imagerie de perfusion dynamique et statique



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Modalités d'Imagerie

Imagerie de perfusion statique

Stress

Rest

Stress

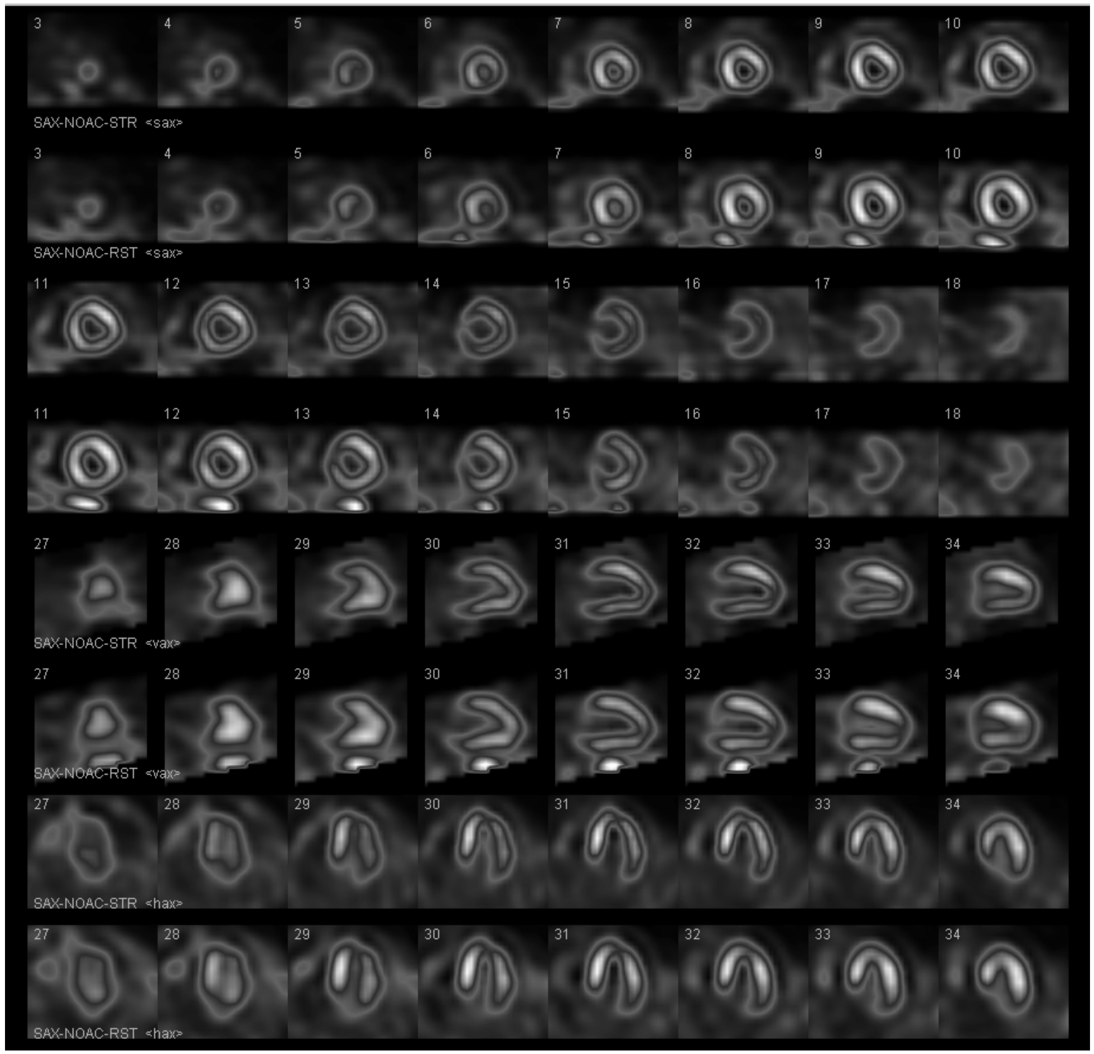
Rest

Stress

Rest

Stress

Rest



SA

SA

VLA

HLA

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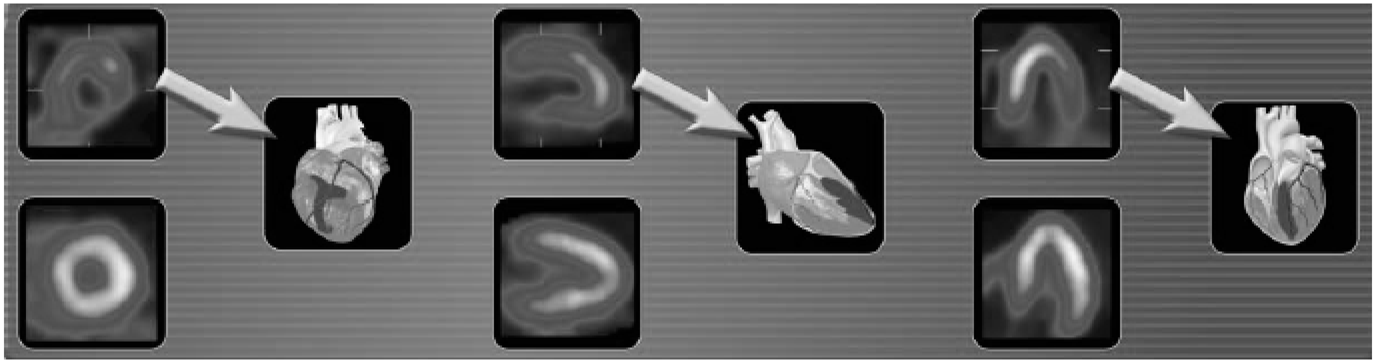
Modalités d'Imagerie

- Imagerie nucléaire: statique (perfusion)

Examens d'efforts (stress exams)

Flux sanguin anormal

Flux sanguin normal



Stress

Rest

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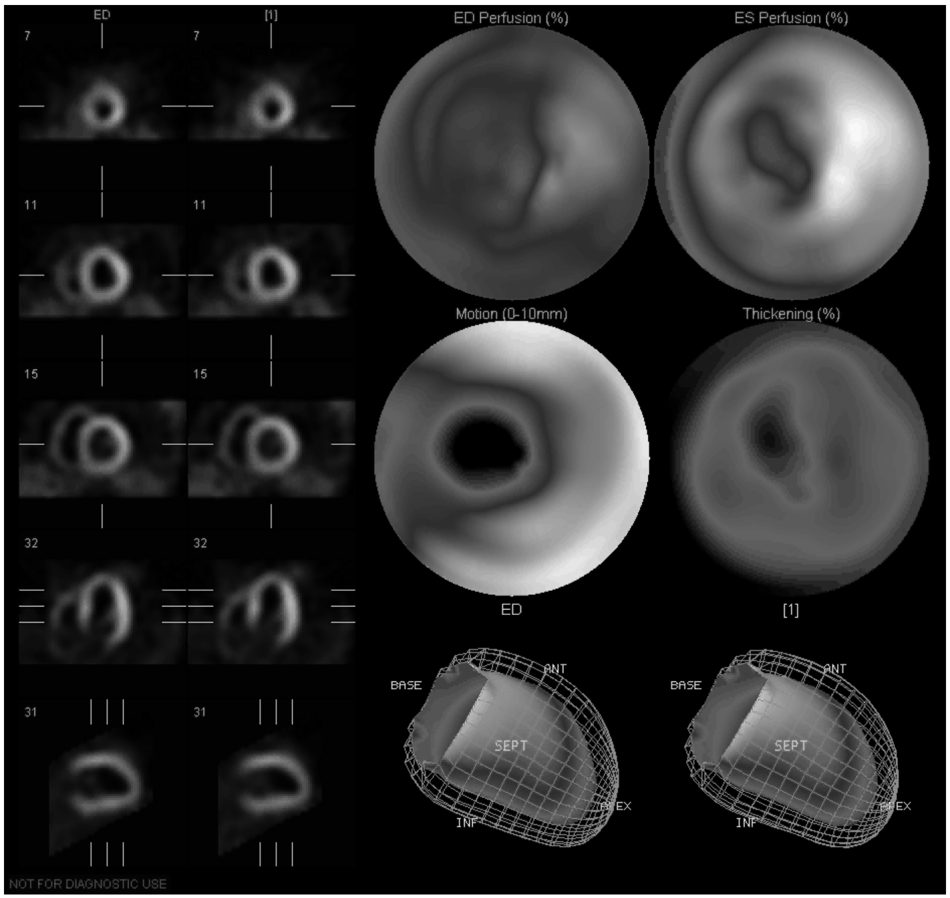
Modalités d'Imagerie

- Imagerie nucléaire: dynamique (perfusion)

SA

HLA

VLA



- Synchronisée par ECG.
- Estimation de la fonction cardiaque:
  - Mouvement: contraction uniforme du myocarde?
  - Fraction d'ejection.

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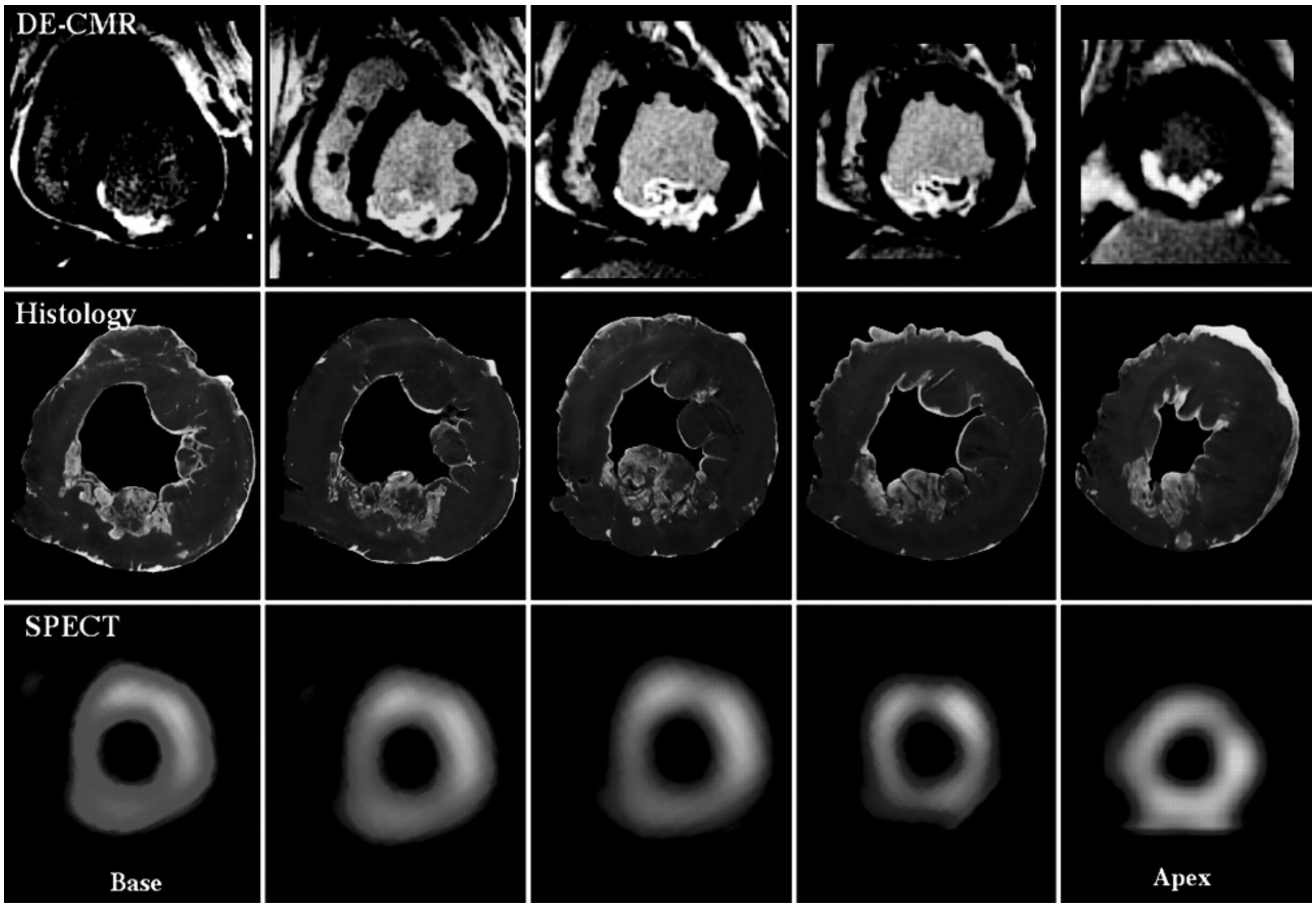
Modalités d'Imagerie

- Nucléaire



- Nucléaire

Review

Economic and biological costs of cardiac imaging

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* Corresponding author

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Cardiovascular Ultrasound 2005, 3:13 doi:10.1186/1476-7120-3-13

Received: 16 May 2005
Accepted: 25 May 2005

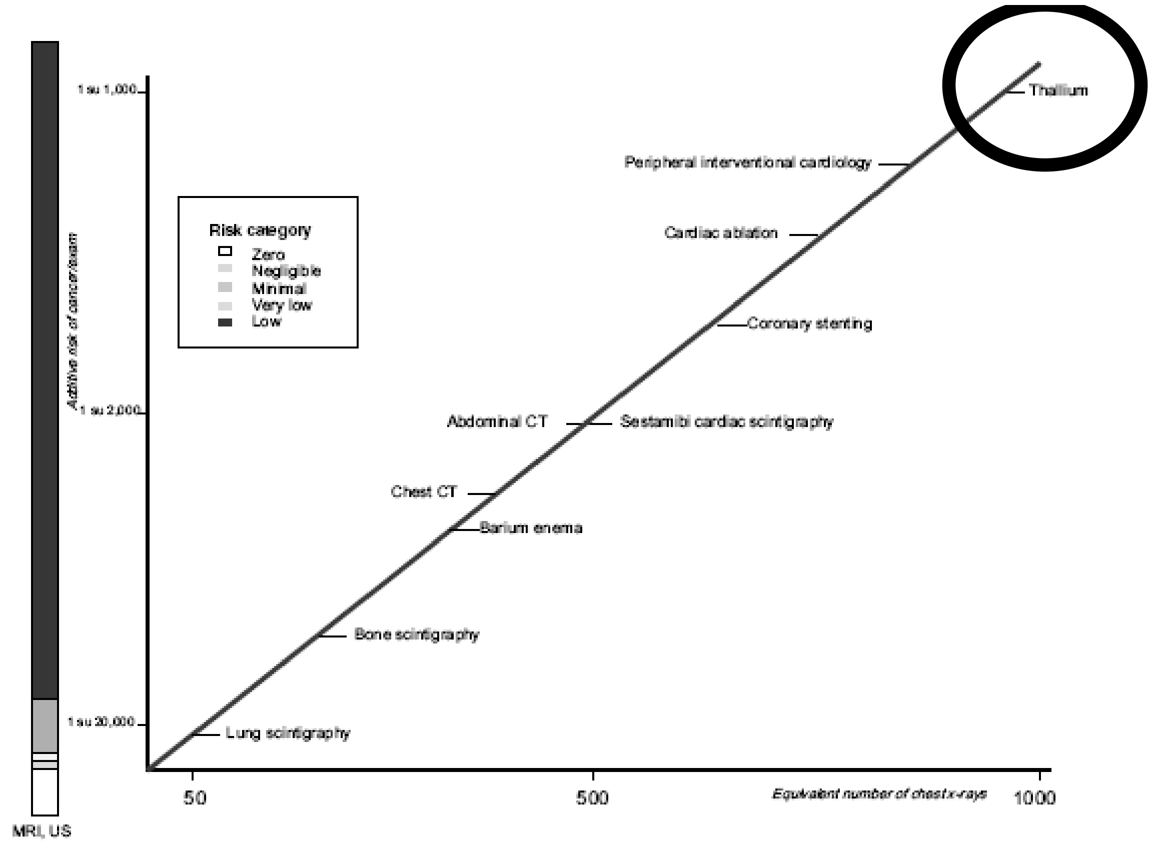

Figure 2
Presentation of cancer risk and radiation dose (in multiples of dose from a simple chest x rays) for some common radiological and nuclear medicine examinations (Modified from Picano E, ref. 37)

Table 1: Radiation doses and estimated cancer risk from common radiological examinations and isotope scans

Type of test	Effective radiation dose (mSv)	Equivalent period of natural background radiation	Lifetime additional risk of cancer/examination	Lost life expectancy	Equivalent n. of chest x-rays
Chest radiograph	0.01	A few days	Negligible risk	2 minutes	1
Skull radiograph	0.1	A few weeks	Minimal risk (1 in 100,000 to 1 in1,000,000)	20 minutes	5
Lung isotope scan	1	A few months to a year	Very low risk (1 in 10,000 to 1 in 100,000)	3 hrs	50
Cardiac gated study	10	A few years (4 years)	Low risk (1 in 2,000)	2 days	500
Thallium scan	20	(8 years)	(1 in 1,000)	4 days	1000

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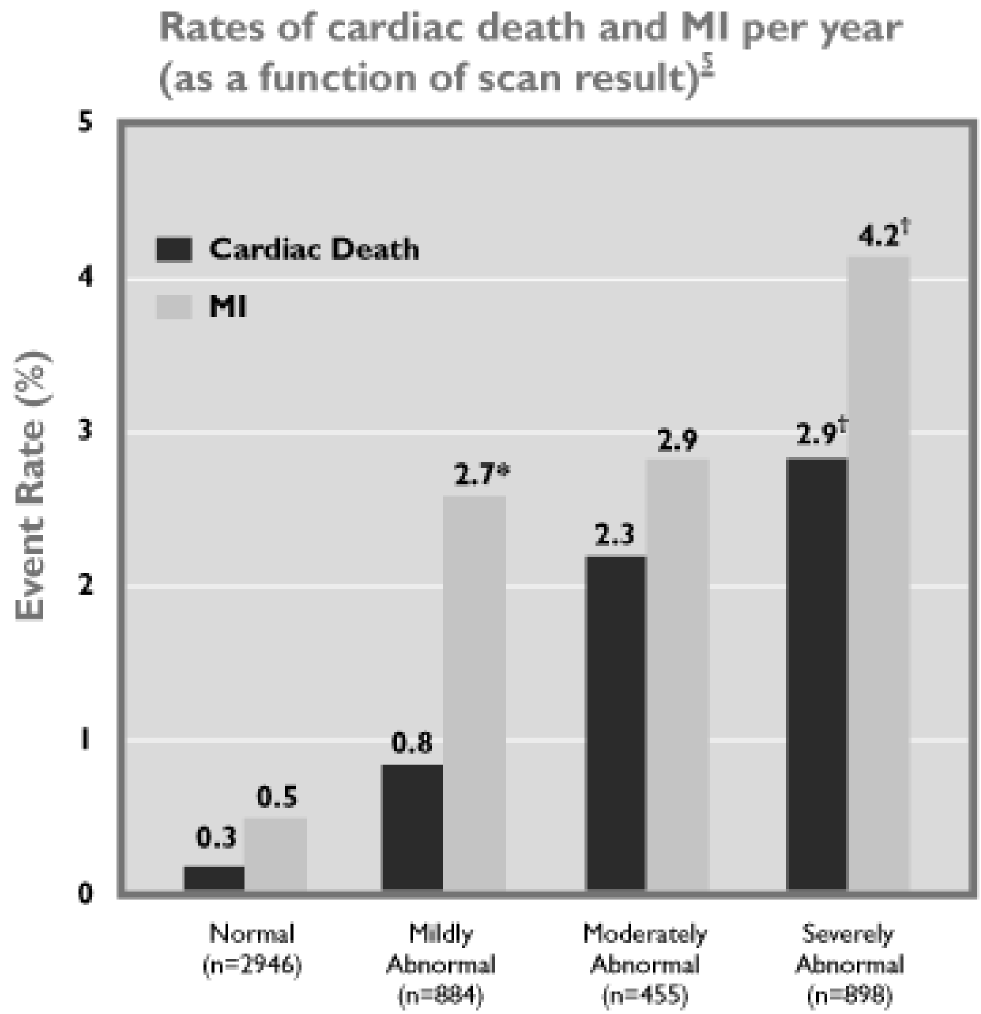
Cardiovascular Ultrasound

BioMed Central

Applications Cliniques

- Nucléaire
  - Bénéfices versus irradiation?
  - Eviter la cathérisation



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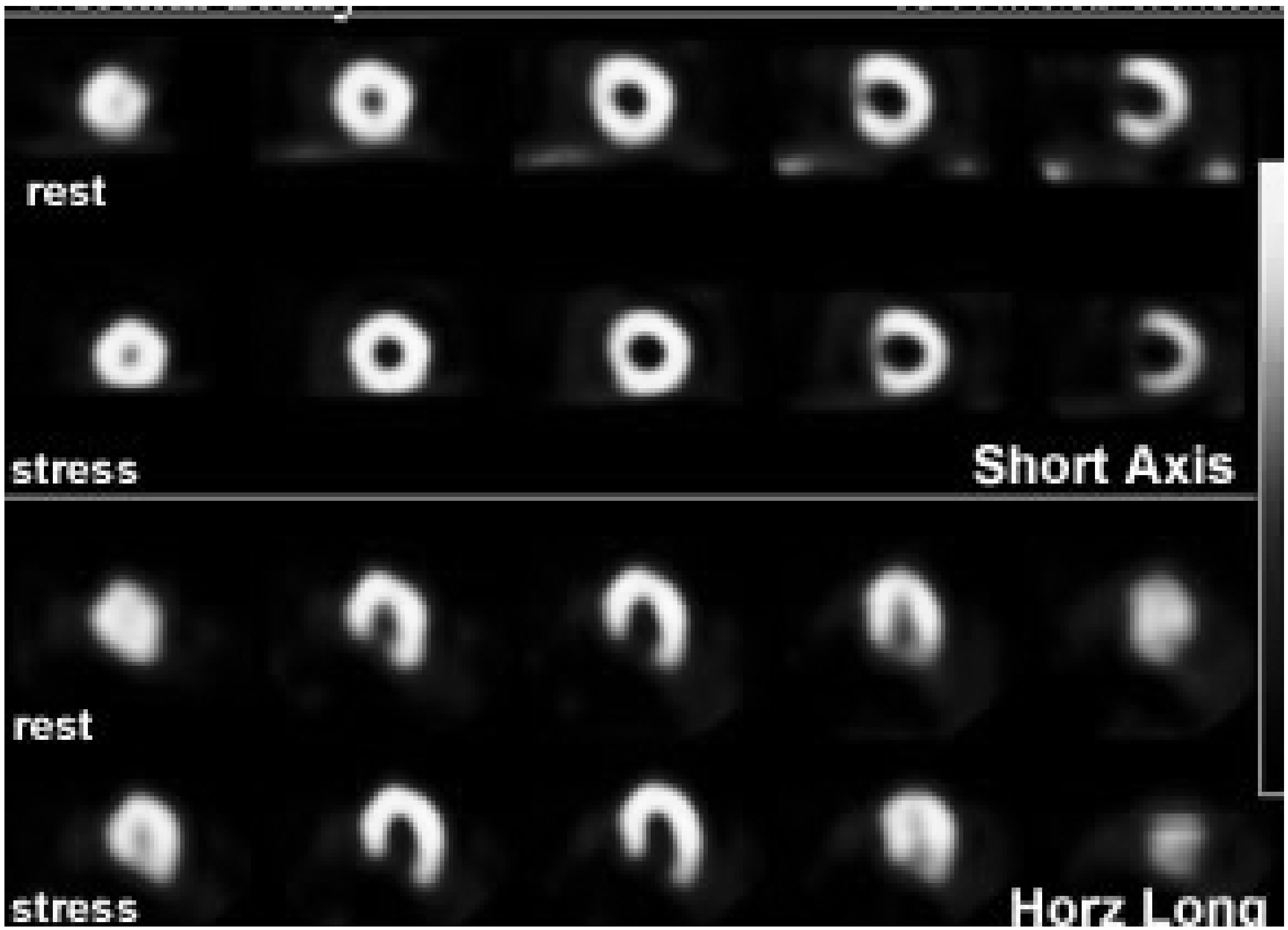
Applications Cliniques

- Nucléaire

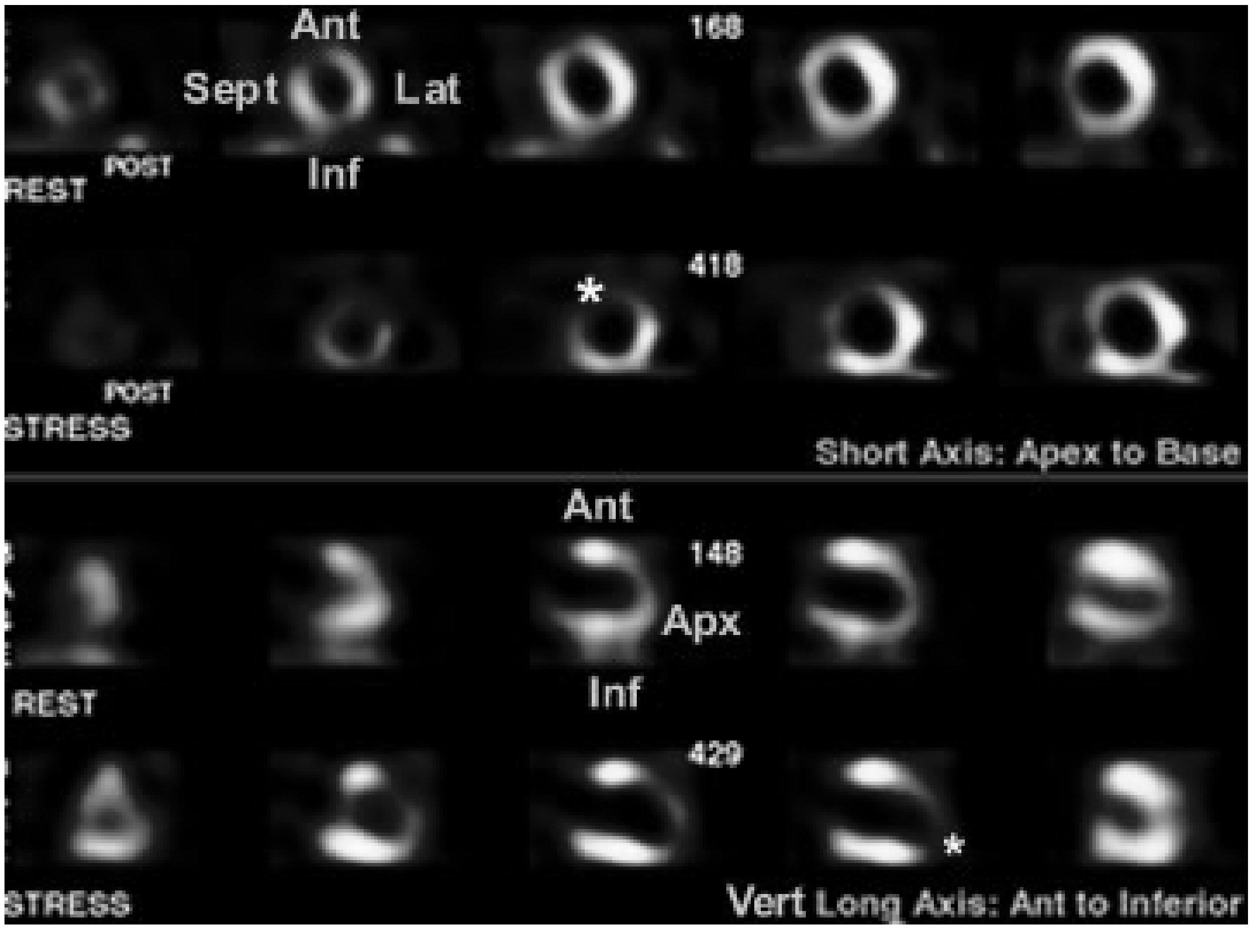
Normal



Applications Cliniques

- Nucléaire

Ischémie



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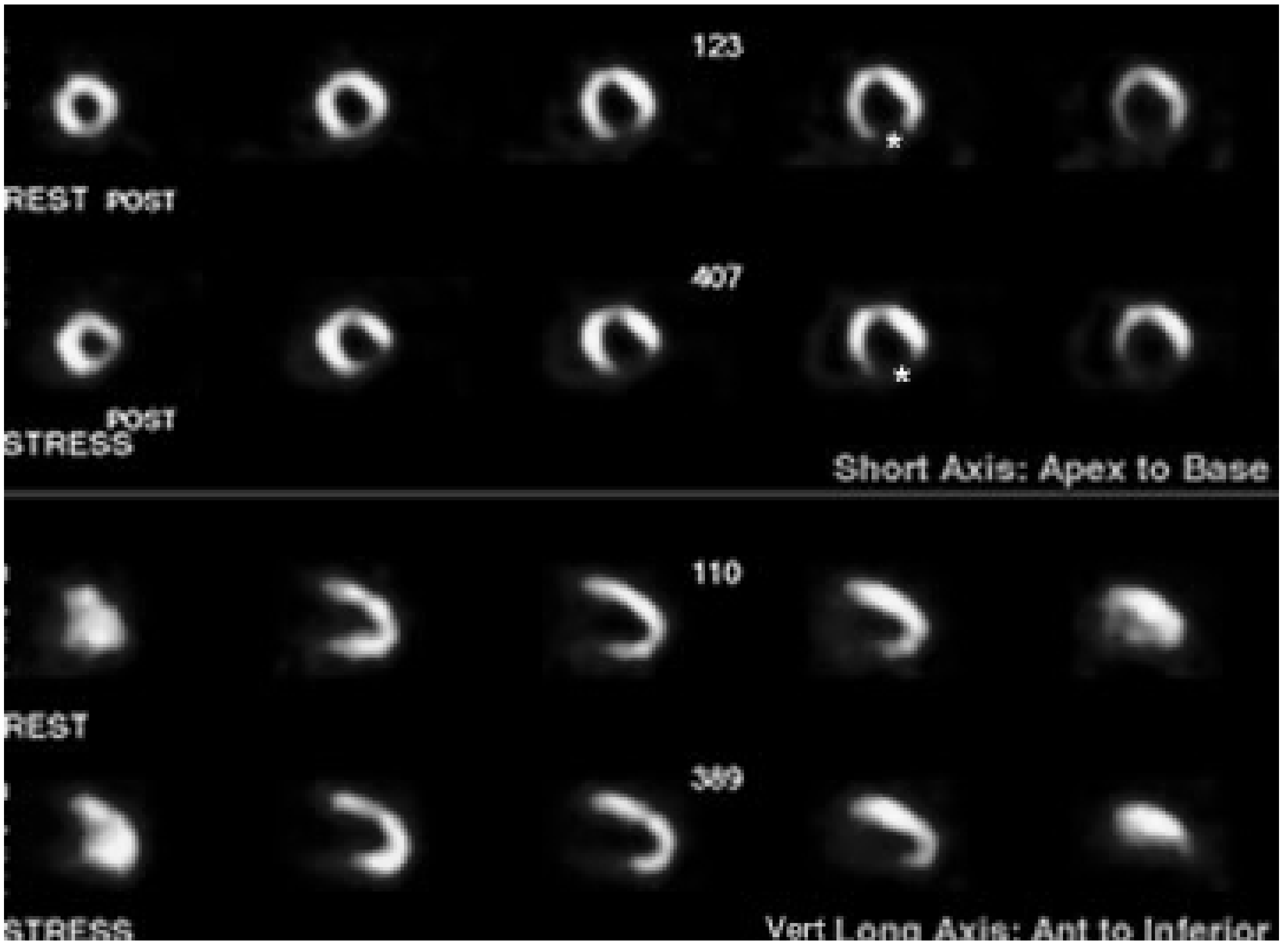
Applications Cliniques

- Nucléaire

Infarctus



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Applications Cliniques

Nucléaire
& IRM

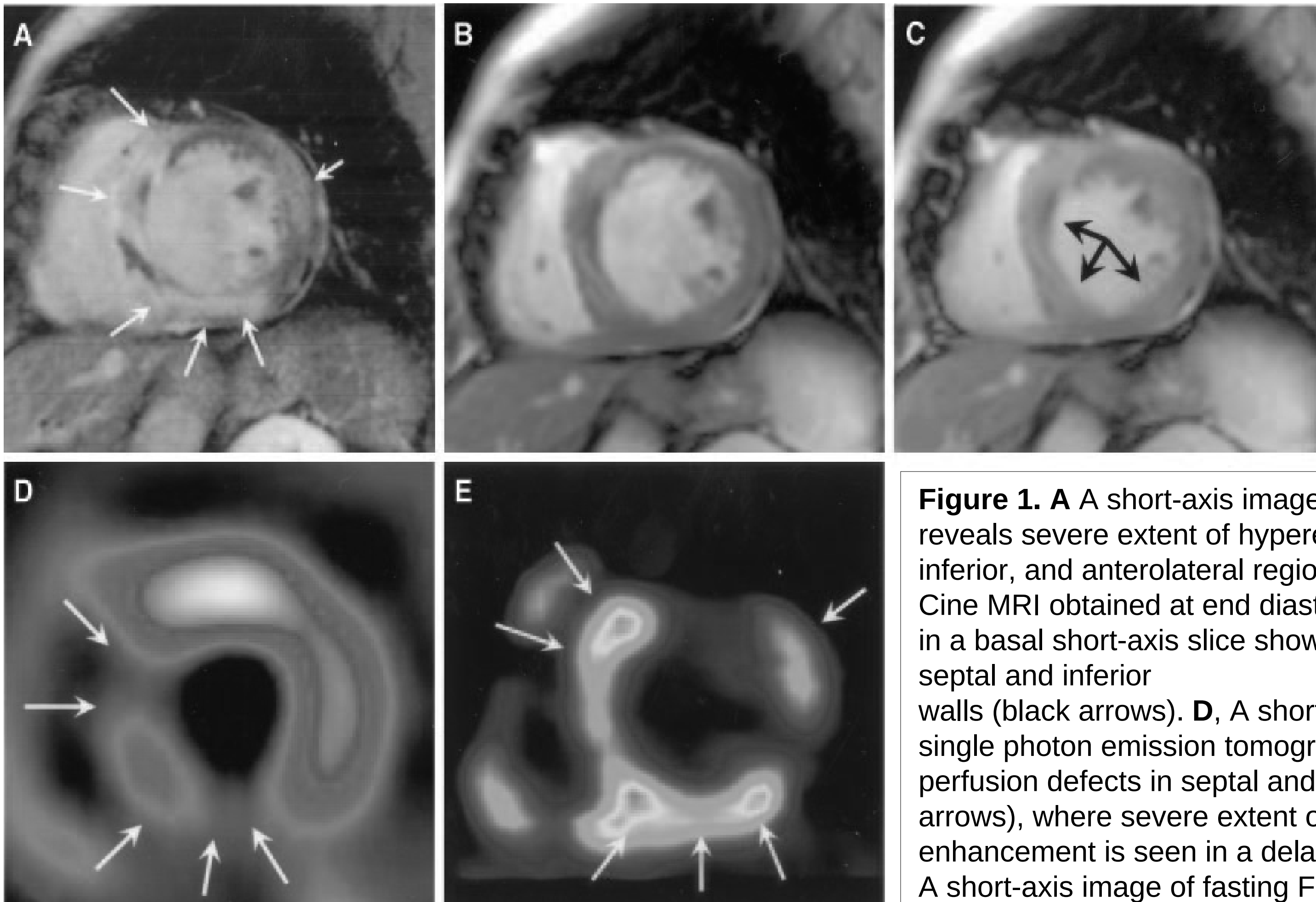

Figure 1. A A short-axis image of delayed-enhanced MRI reveals severe extent of hyperenhancement in septal, inferior, and anterolateral regions (white arrows). **B and C**, Cine MRI obtained at end diastole (B) and end systole (C) in a basal short-axis slice show wall motion abnormalities in septal and inferior walls (black arrows). **D**, A short-axis image of 201thallium single photon emission tomography demonstrates perfusion defects in septal and inferior regions (white arrows), where severe extent of enhancement is seen in a delayed enhanced MR image. **E**, A short-axis image of fasting FDG PET shows FDG accumulation in septal, inferior, and anterolateral walls (white arrows), corresponding to areas with late enhancement.

Plan

3 Applications Cliniques

et ….applications en recherche

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Applications Cliniques

Mesures Principales:

- **Cœur:**
  - **Volumes** du ventricule gauche à fin-systole (ESv) et fin diastole (EDv) et la **fraction d'éjection:** EF = (EDv-ESv)/EDv
  - Masse du myocarde
  - Épaississement du myocarde
- **Vaisseaux:**
  - Dimension des plaques.
  - **Score de calcium** des plaques.

Applications Cliniques

Mesures du volume **cardiaque**

Sur les images nucléaires synchronisées:

1. Injection d'un radiotraceur (ex: technetium-99 (½ vie ~6 heures)).
2. Détection des émissions γ autour du ventricule cardiaque.
   - Synchronisation des périodes de comptage par l'ECG
   - Comparaison des comptages à fin-diastole & fin-systole.
3. Ratio ~ EDV/ESV = Fraction d'éjection

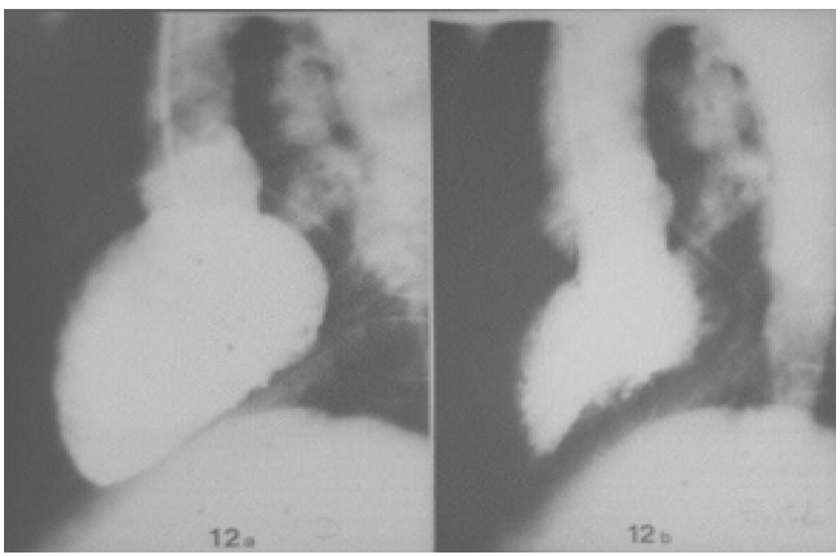
$\Rightarrow$ Pas quantitatif

Applications Cliniques

Mesures du volume cardiaque

Sur les images d'angiographie RX:

1. Injection d'un agent de contraste radio-opaque.
2. Acquisition de série d'images rayons-X.
3. Estimation des volumes sur images 2D.



⇒ très approximatif

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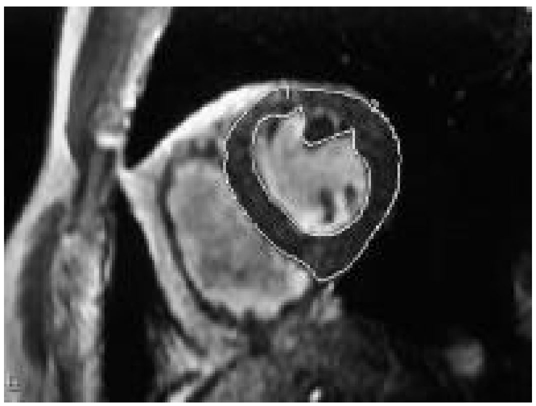
Applications Cliniques

Mesures du volume cardiaque

Sur les images IRM:

1. Contours de l'endocarde sur des images représentant des mesures de densité de protons.
2. Estimation des volumes sur une série de coupes.

⇒ Acquisition lente (synchronisée) et résolution spatiale limitée.



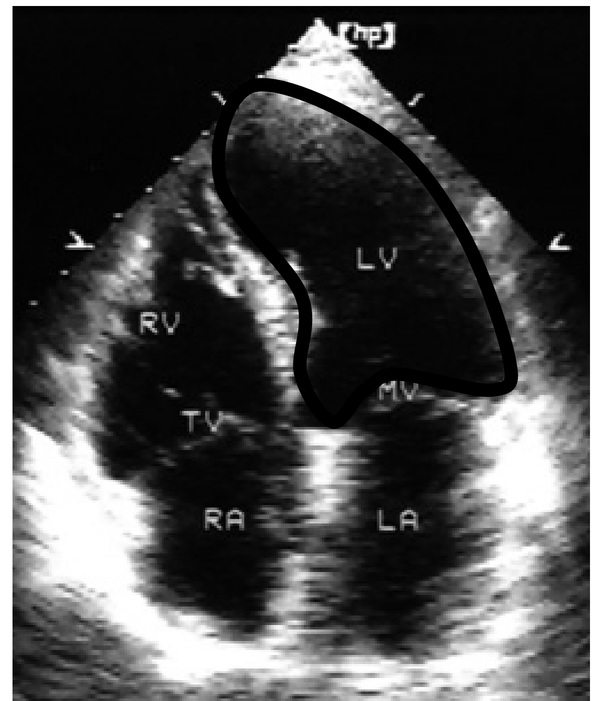
Source: www.emedecine.com

Applications Cliniques

Mesures du volume cardiaque

Sur les images d' échocardiographie:

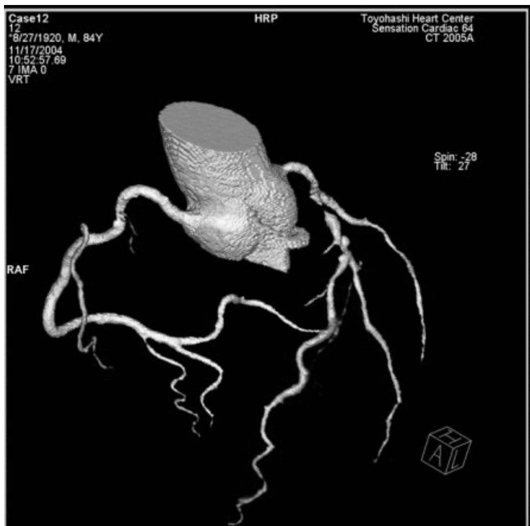
1. Acquisitions 2D (parasternales, apicales ou trans-oesophageales)
2. Contours de l'endocarde et de l'épicarde.



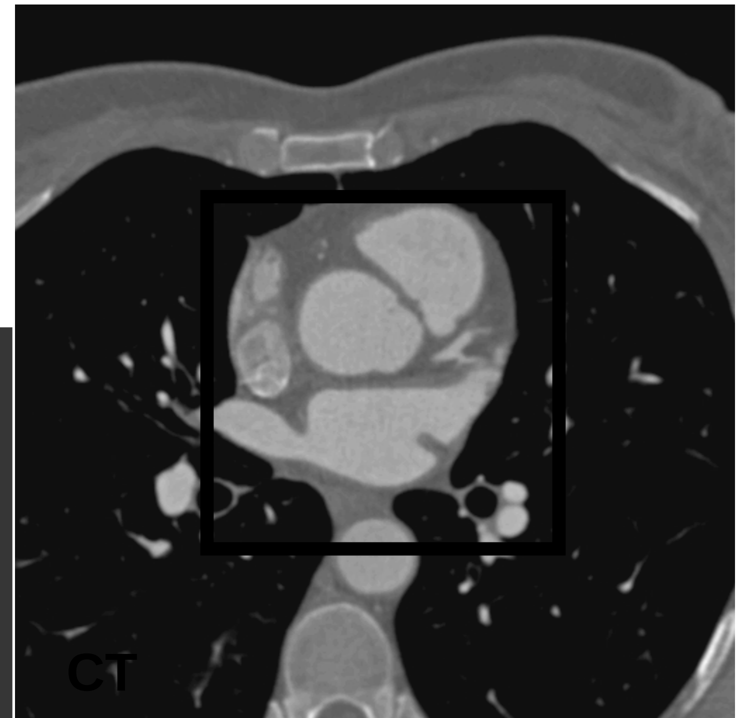
⇒ Difficultés de traitements automatiques de segmentation

Applications Cliniques

Score de Calcification



Sur CT



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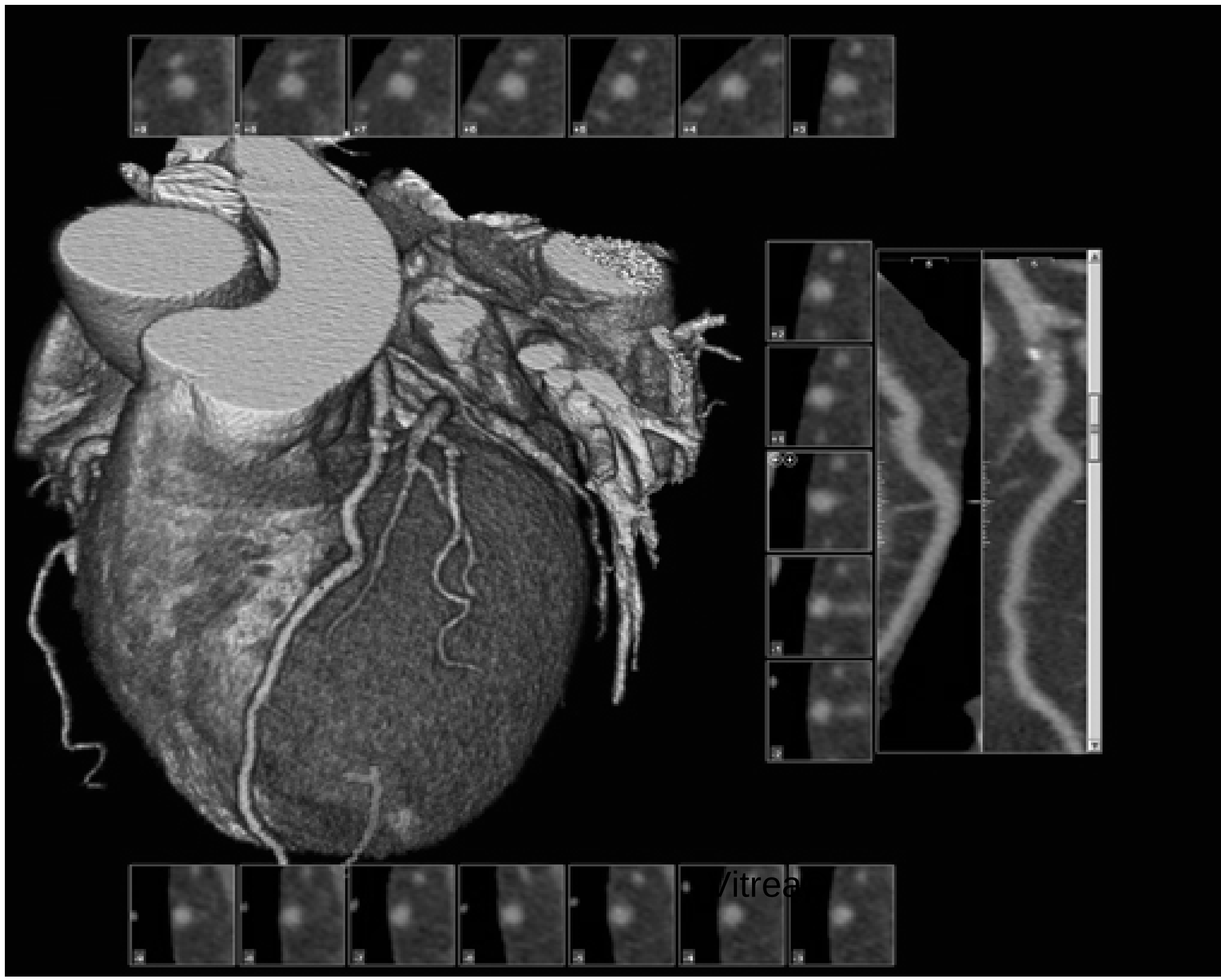
Applications Cliniques

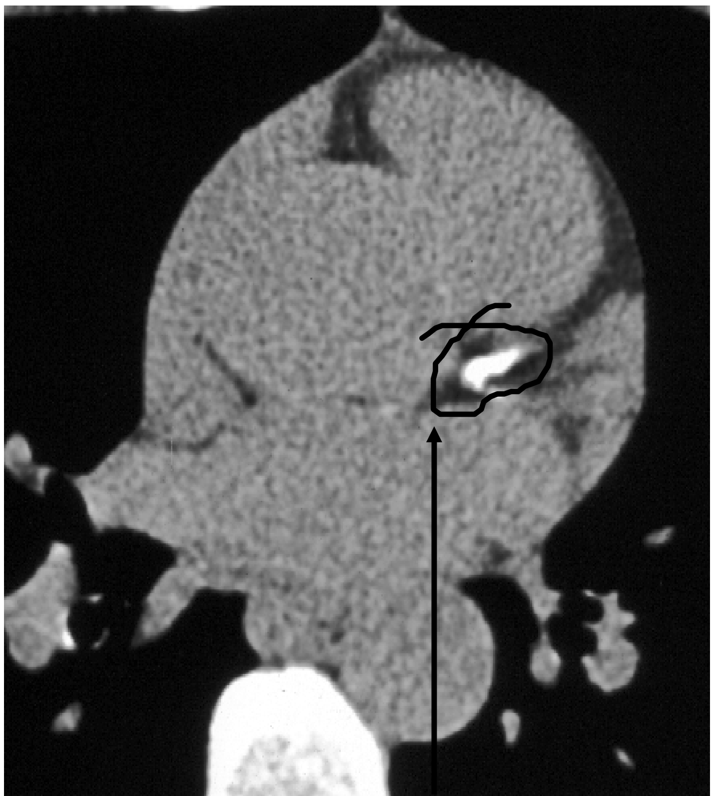
Score de Calcification

Sur CT



Applications Cliniques

Score de Calcification



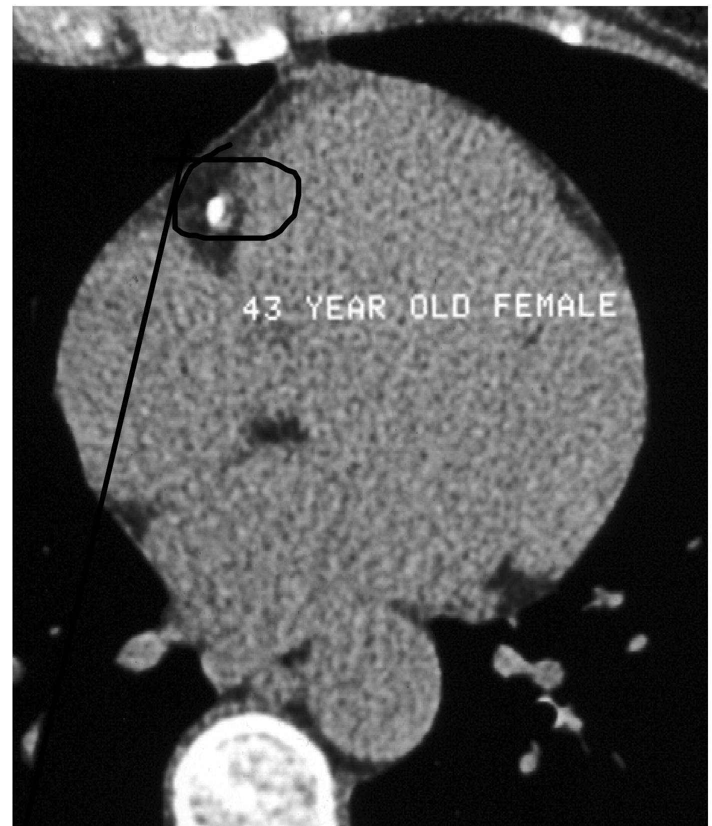
Area = 15 mm^2
Peak CT = 450
Score = 15 x 4 = 60

Sur CT

Total Score = S

Hn x-factor
(Agatston Scoring)

130-199	**1**
200-299	**2**
300-399	**3**
>400	**4**



Area = 8 mm^2
Peak CT = 290
Score = 8 x 2 = 16

Applications Recherche

- Modélisation cardiaque

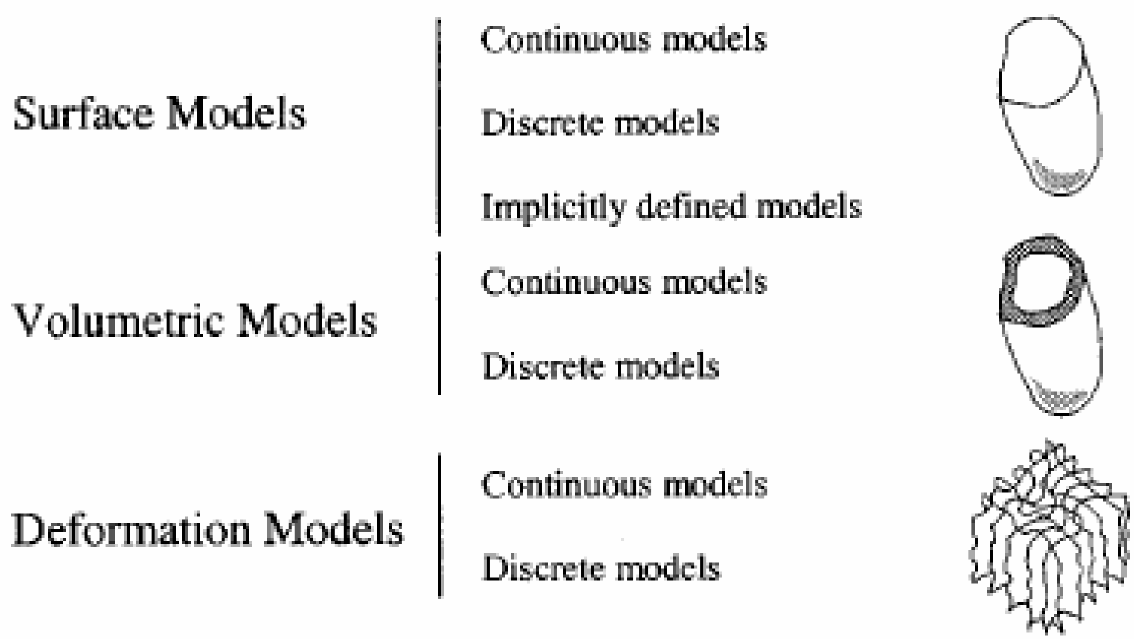
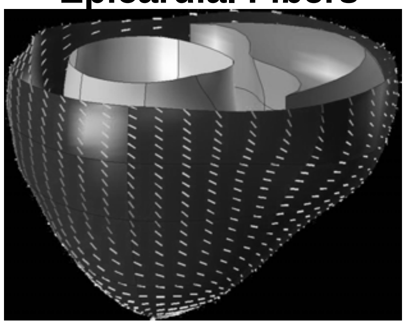


Fig. 2. Proposed classification of cardiac modeling approaches.

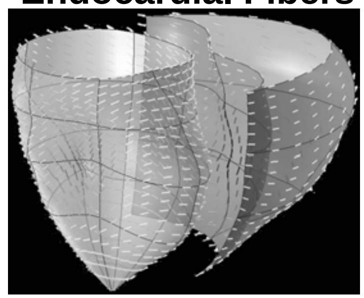
Applications Recherche

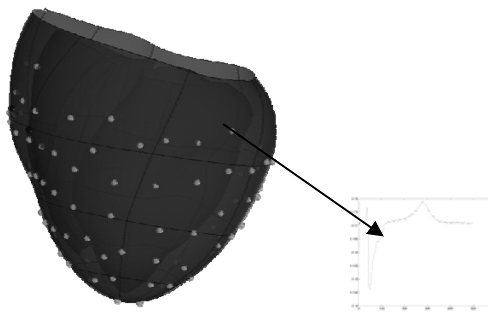
Modèle Cardiaque: FEM

Epicardial Fibers



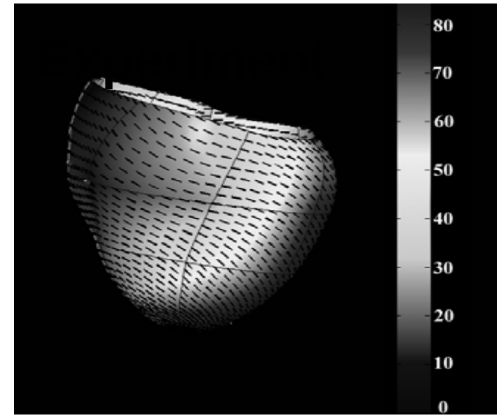
Endocardial Fibers

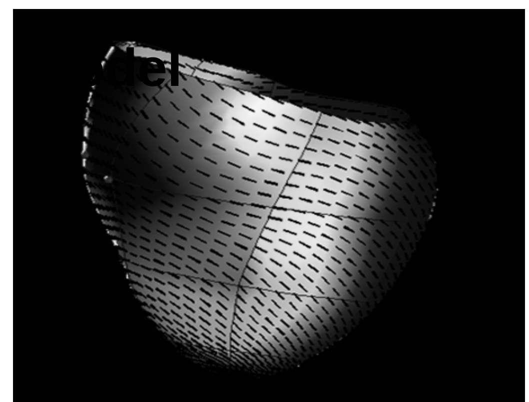




Physiologie musculaire

Physiologie électrique





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- IRM

Kinematics of the Heart: Strain-Rate Imaging From Time-Resolved Three-Dimensional Phase Contrast MRI

Pernilla Selskog*, Einar Heiberg, Tino Ebbers, Lars Wigström, and Matts Karlsson

A. Theory

To calculate strain-rate from a velocity field, the 3×3 velocity gradient tensor (Jacobian) L_{ij} is calculated according to

$$L_{ij} = \frac{\partial u_i}{\partial x_j} \quad (1)$$

where u_i, $i = 1, 2, 3$ are the three velocity components in the x_j direction, $j = 1, 2, 3$ [7].

Strain-rate is represented by the strain-rate tensor D_{ij}, which is the symmetric part of L_{ij} [7]

$$D_{ij} = \frac{1}{2}\left(\frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i}\right). \quad (2)$$

The eigenvalues and eigenvectors of the strain-rate tensor are the principal values and the principal directions of strain-rate in the myocardium. The sign of the eigenvalue distinguishes between positive and negative material stretching in the direction of the corresponding eigenvector.

An invariant I_t representing the total amount of strain-rate in each voxel was calculated according to

$$I_t = \lambda_1^2 + \lambda_2^2 + \lambda_3^2. \quad (3)$$

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D. In Vivo Data

Calculations were performed using velocity data for a healthy 24-year-old volunteer with a heart rate of 66 bpm. The velocity measurements were performed using a 1.5 T Signa Horizon Echospeed scanner (General Electric Medical Systems, Milwaukee, WI) with a 3-D cine phase contrast pulse sequence, allowing offline retrospective gating based on the signal from a pulse oximeter [9], [12].

Velocity encoded data were acquired in all three directions at 12 times for each phase-encoding step. A complete set of 3-D k-space data was interpolated in the time domain to 32 time-frames using an interpolation method based on a normalized convolution algorithm with a Gaussian interpolation function [9]. Using peripheral gating, i.e., gating based on the signal from a pulse oximeter on the finger, the first time frame corresponds to approximately 200 ms after onset of electrical systole. Only a cylindrical region of k-space was collected in order to reduce the acquisition time by not acquiring the corners of the $k_y - k_z$ plane [13].

All three velocity components were measured in a $30.0 \times 30.0 \times 11.2$ cm axial volume, suitable for myocardial motion studies, with a spatial resolution of $1.2 \times 4.0 \times 4.0$ mm and a temporal resolution of 108 ms (TR = 27 ms, TE = 7.2 ms, VENC = 0.18 m/s, flip angle = 20°, one NEX, acquisition time 39:14 min). The technique includes automated unwrapping of aliased velocities [9]. The MRI data were resampled using zero-filling in the Fourier domain to a voxel size of $1.5 \times 1.5 \times 1.5$ mm. Saturation bands superior and inferior to the imaging volume were used to reduce the signal from the blood [14]. After the 3-D inverse Fourier transform, the phase contribution from concomitant field (Maxwell) effects was calculated and subtracted from the velocity data [15].

A linear 3-D least square fit to the data from regions containing stationary tissue was subtracted for correction of eddy current effects [16]. Conventional 2-D time-resolved gradient-echo images were acquired in short- and long-axis cardiac views for better depiction of anatomical landmarks.

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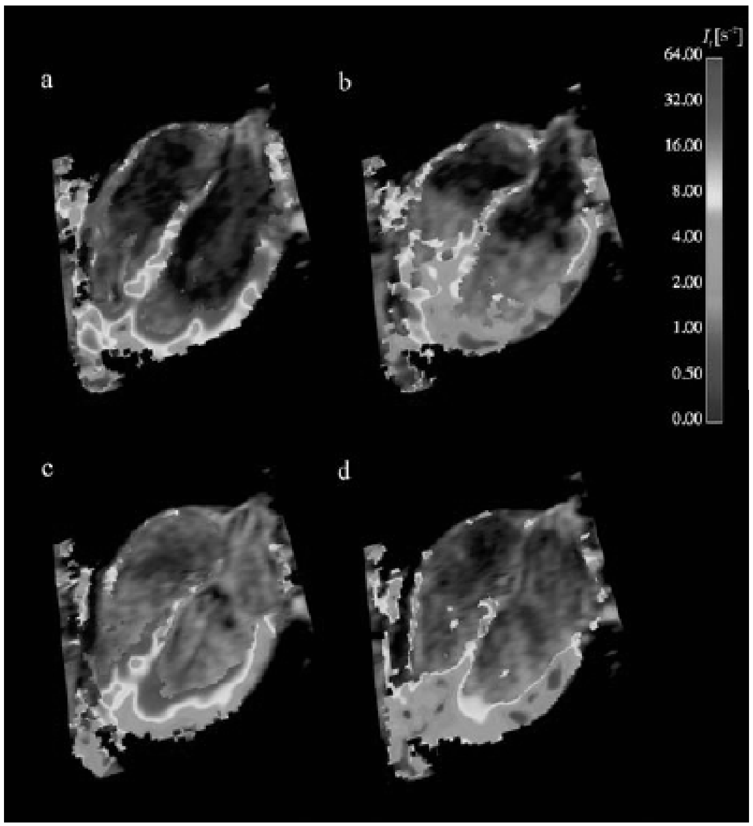

Fig. 4. A color map of the invariant (I_t) for a long-axis slice through the myocardium in (a) early diastole, (b) end diastole, (c) systole, and (d) end systole. A time plot of I_t at a point in the lateral left ventricular wall is shown in Fig. 5.

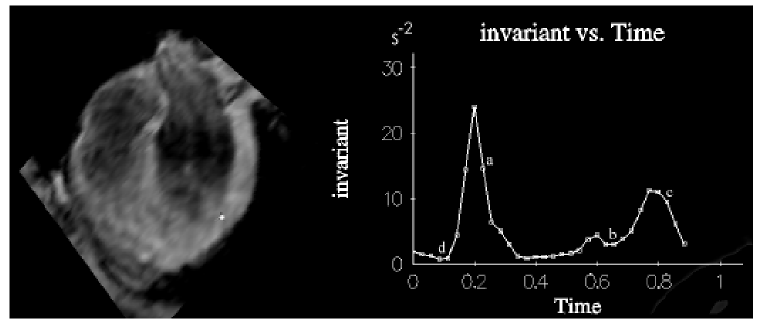

Fig. 5. Strain-rate invariant $I_t = \lambda_1^2 + \lambda_2^2 + \lambda_3^2$ at a point in the lateral left ventricular wall (highlighted in the left figure) throughout the cardiac cycle (right). Peaks are found in early diastole and in systole. (a)–(d) refer to the time frames displayed in Fig. 4.

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Applications Recherche

- Ultrasons

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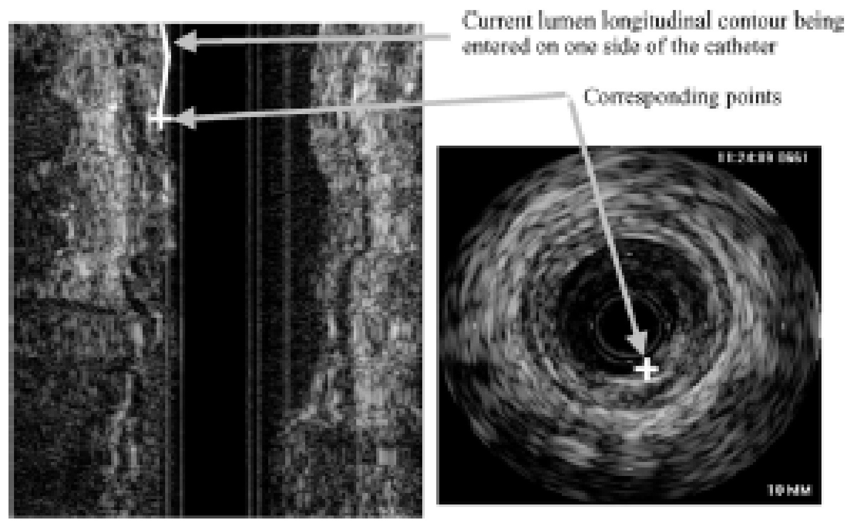
Intravascular Ultrasound Image Segmentation: A Three-Dimensional Fast-Marching Method Based on Gray Level Distributions

Marie-Hélène Roy Cardinal, Jean Meunier, Gilles Soulez, Roch L. Maurice, Éric Therasse, and Guy Cloutier*



(a) (b)

Fig. 2. (a) Longitudinal IVUS image of the 3-D data set in which a contour is being entered and (b) current longitudinal contour point displayed on the corresponding 2-D IVUS image. At the end of the initialization procedure, 4 lines were traced in 3 longitudinal images (that converts into 6 points on the lumen border and 6 points inside the media in each axial frame).

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The Rayleigh PDF $p_X(x)$ with parameter a^2, where X is the gray level taking values in $[1, \ldots, 256]$, is given by

$$p_X(x; a^2) = \frac{x}{a^2} \exp\left(-\frac{x^2}{2a^2}\right) \quad (1)$$

with x, $a^2 > 0$ and the variance $\sigma^2 = a^2(4 - \pi)/2$.

IVUS data, in which there are M different tissue structures, were modeled by a mixture of M Rayleigh PDFs with parameters $\Theta = \{(\omega_j, a_j^2)\}_{j=1}^M$, where ω_j is the proportion of the jth component of the mixture so that $\sum_{j=1}^M \omega_j = 1$. The PDF mixture then becomes

$$p_{X\,\Theta}(x|\Theta) = \sum_{j=1}^{M} \omega_j p\left(x|a_j^2\right). \quad (2)$$

To describe the global data PDF, the parameters (ω_j, a_j^2) of each distribution composing the mixture need to be estimated. The EM algorithm is necessary because $\hat{\Theta}$, the mixture parameter maximizing the likelihood of $p(X|\Theta)$, cannot be solved analytically. A hidden variable Y, the tissue class taking values $[1, \ldots, M]$, must be introduced at this point. The log-likelihood of the joint distribution of $(X, Y) = \{(x_i, y_i)\}_{i=1}^N$, where N is the data size, is

$$\log\left(p_{X,Y|\Theta}(x, y\,\Theta)\right) = \sum_{i=1}^{N} \log p(y_i) p(x_i\, y_i, \Theta). \quad (3)$$

The first step of the EM algorithm (Expectation) is the calculation of the cost function $Q(\Theta, \Theta') = E_Y[\log(P(X, Y\,\Theta))|X, \Theta']$, the expected value of the log-likelihood of (X, Y), the joint distribution, given the observed data X and $\Theta' = \{(\omega_j', a_j^{2\prime})\}_{j=1}^M$, a previous estimate of the mixture parameters. The next step is to evaluate $\hat{\Theta}$, the new parameter estimate, by maximizing $Q(\Theta, \Theta')$ with respect to Θ, that can now be done analytically.

TABLE I
ESTIMATED PROBABILITY DENSITY FUNCTION PARAMETERS FOR 30 RUNS OF THE EM ALGORITHM ON 1 IVUS PULLBACK

Component	ω (%)	a^2
Lumen	29.63±0.88	0.6456±0.0014
Intima and Plaque	20.72±0.72	341.53±6.49
Media	13.49±0.11	22.43±0.39
Surrounding Tissues	36.19±0.24	2283.04±12.09

$$F_m(i,j,k) = \left(1 + \frac{1}{N_\nu} \sum_{s \in \nu} \frac{logP_m(I_s)}{\frac{1}{N_L - 1} \sum_{l \neq m, l \in L} logP_l(I_s)}\right)^{-1}. \quad (10)$$

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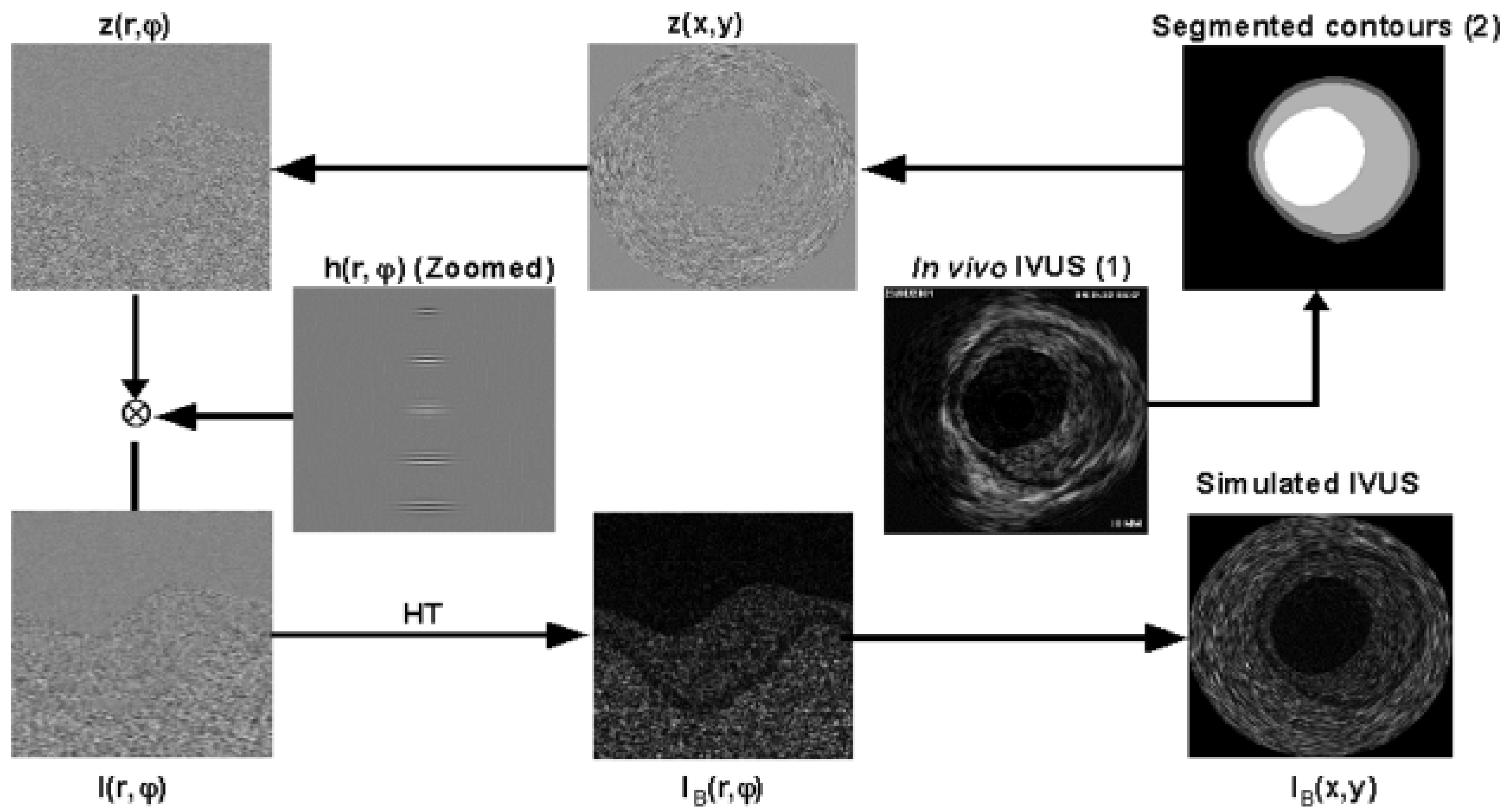

Fig. 3. Schematic implementation of the image-formation model. An *in vivo* IVUS pullback (1) was used to create the vessel geometry (2). $z(x, y)$ is a function representing the acoustic impedance variations; $z(r, \varphi)$ is the acoustic impedance function mapped in polar coordinates; $h(r, \varphi)$ is the polar PSF, with a beam width that increases with depth; $\otimes$ is the 2-D-convolution operator; $I(r, \varphi)$ is the simulated polar radio-frequency image; $I_B(r, \varphi)$ is the polar B-mode image, that was computed using the Hilbert transform (HT) [33] of $I(r, \varphi)$; $I_B(x, y)$ is the Cartesian B-mode image or simulated IVUS image. This simulation strategy was repeated for the whole image series of a pullback within a diseased superficial femoral artery. For more details on the simulation model, refer to [32].

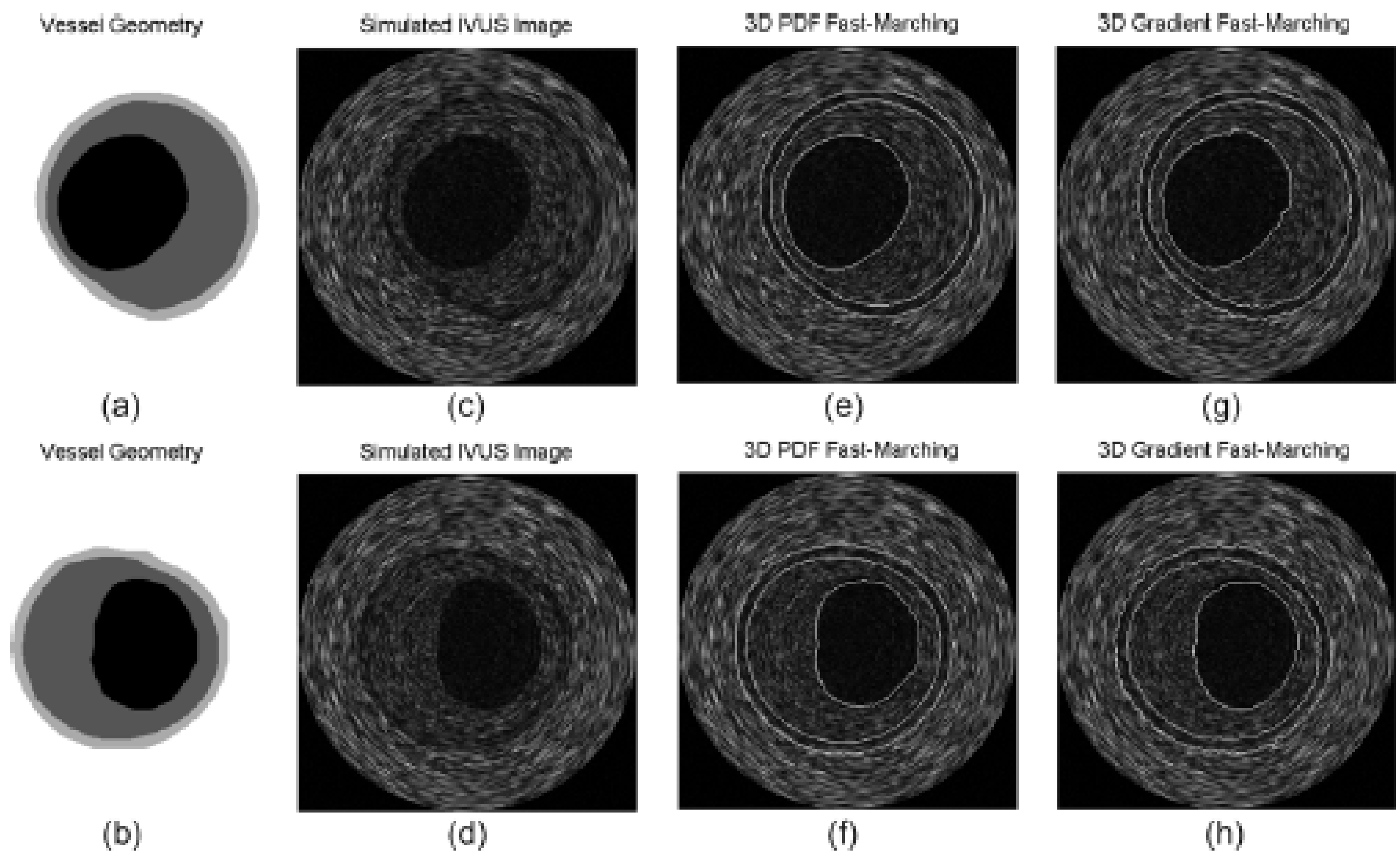

Fig. 5. Segmentation examples on simulated data. (a) and (b) Vessel geometries and (c) and (d) simulated IVUS cross-sectional images. Lumen, thickened intima, and media detected boundaries with (e) and (f) PDFs and (g) and (h) gradient 3-D fast-marching methods.

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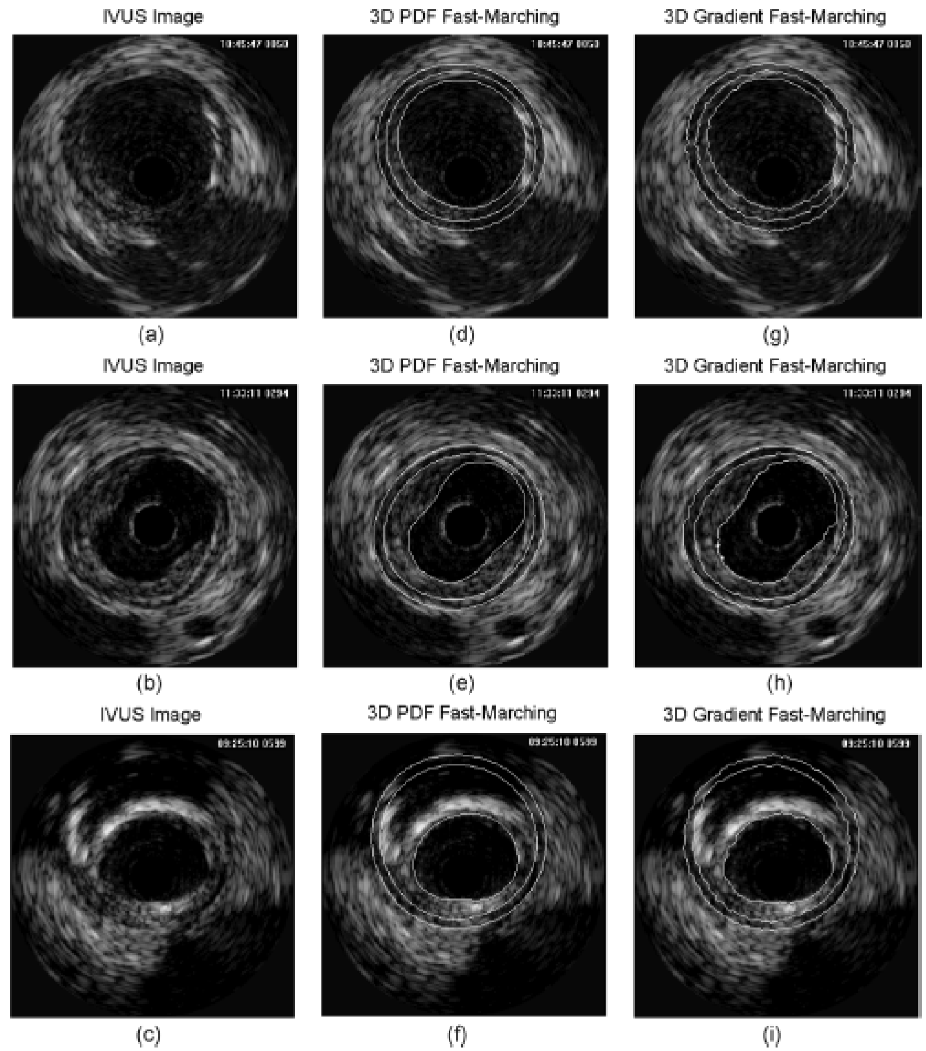
- Ultrasons



6. (a)-(c) Typical IVUS cross-sectional images and corresponding segmentation results. Lumen, thickened intima, and media detected boundaries with (d)-(f) 's and (g)-(i) gradient 3-D fast-marching methods.

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IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING, VOL. 46, NO. 10, OCTOBER 1999

Fusion of Angiography and Intravascular Ultrasound *in vivo*: Establishing the Absolute 3-D Frame Orientation

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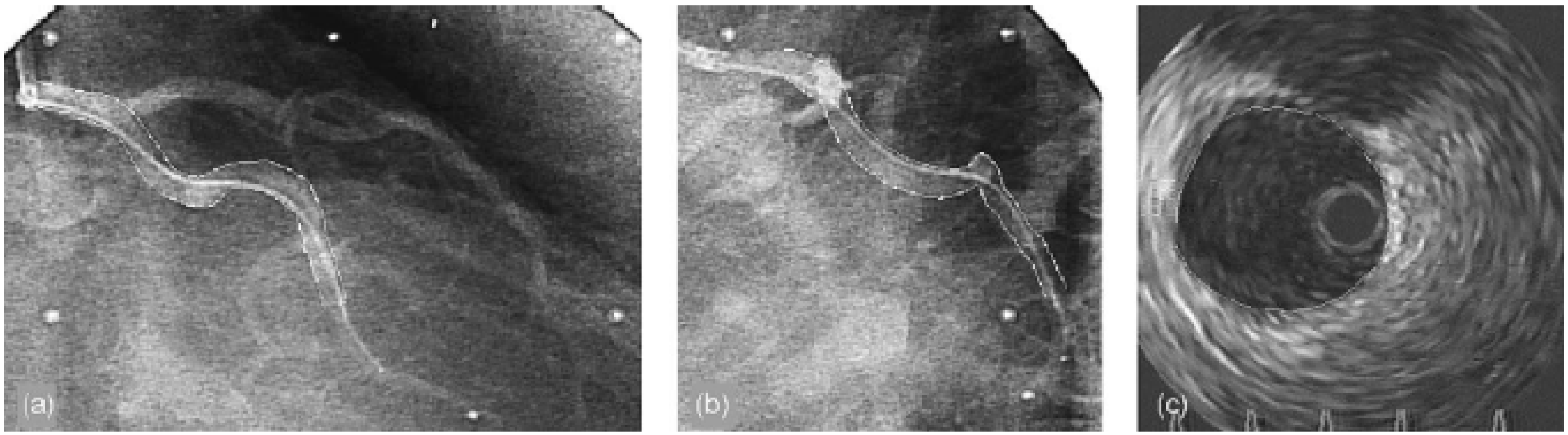

Fig. 1. Angiograms of a circumflex artery in a patient heart using (a) 30° right and (b) 60° left anterior oblique projections, with catheter inserted and the vessel lumen outline detected. (c) Automated detection of the lumen border in one of the IVUS images, note the out-of-center position of the catheter.

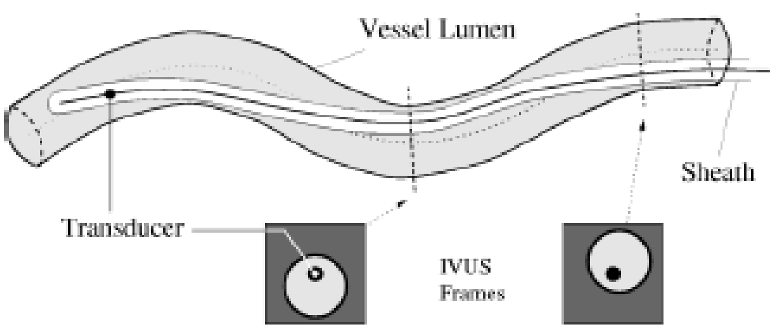

Fig. 2. Bending of the imaging catheter within the vessel due to vessel curvature, seeking position of minimum energy, and associated out-of-center position in IVUS images. While the absolute orientations of the IVUS cross sections in the images are arbitrary, their orientations relative to each other are fixed.

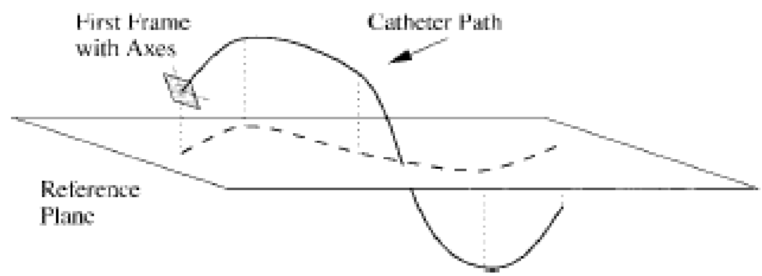

Fig. 3. Reference plane for specification of the initial orientation; the horizontal axis of the first IVUS frame is aligned parallel to the plane.

the next frame is calculated by a discrete version of the Frenet–Serret formulas. This algorithm has been extensively validated in several *in vitro* studies [12], [19].