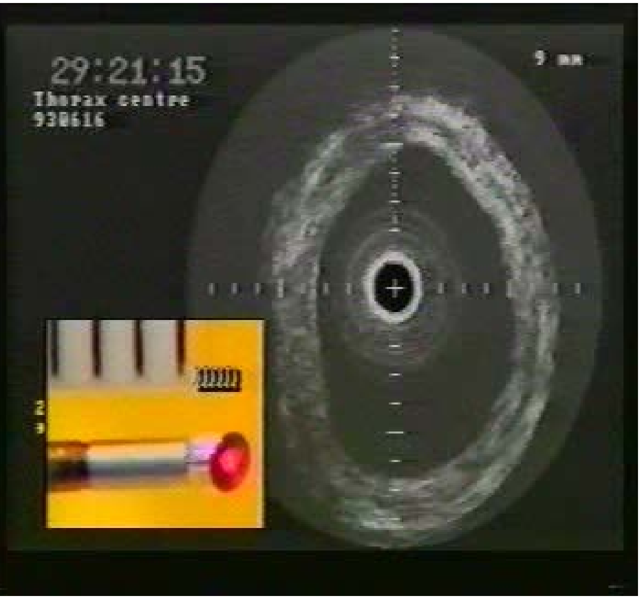
Ultrasound imaging
Intravascular Ultrasound (IVUS)



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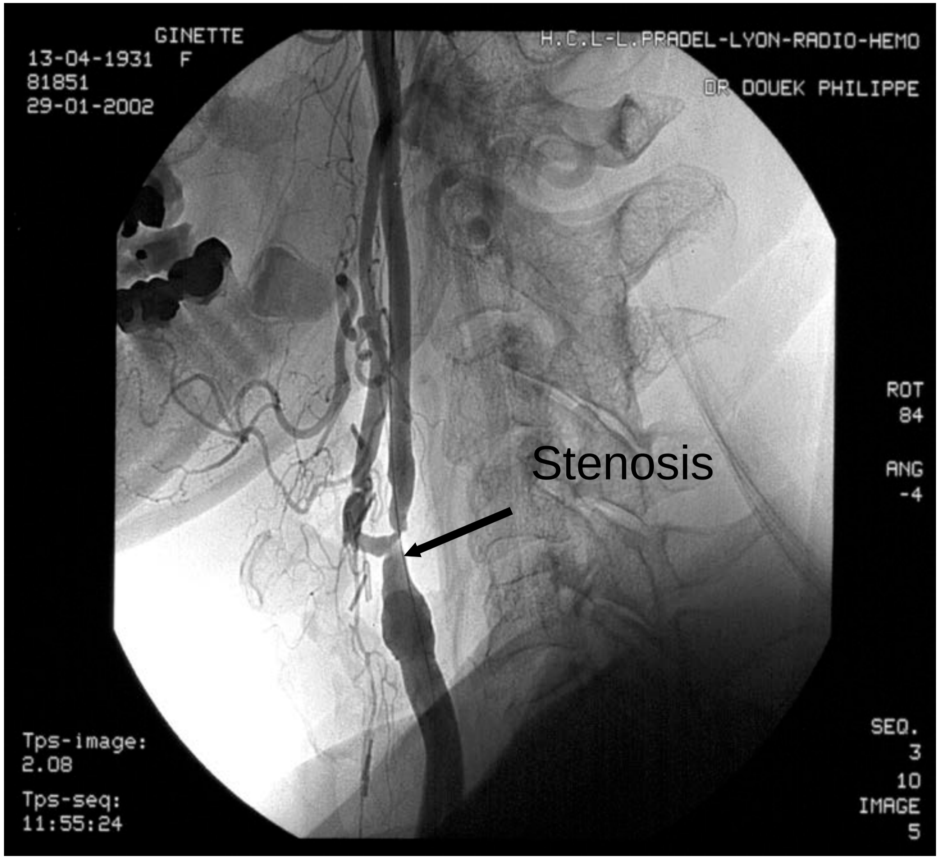
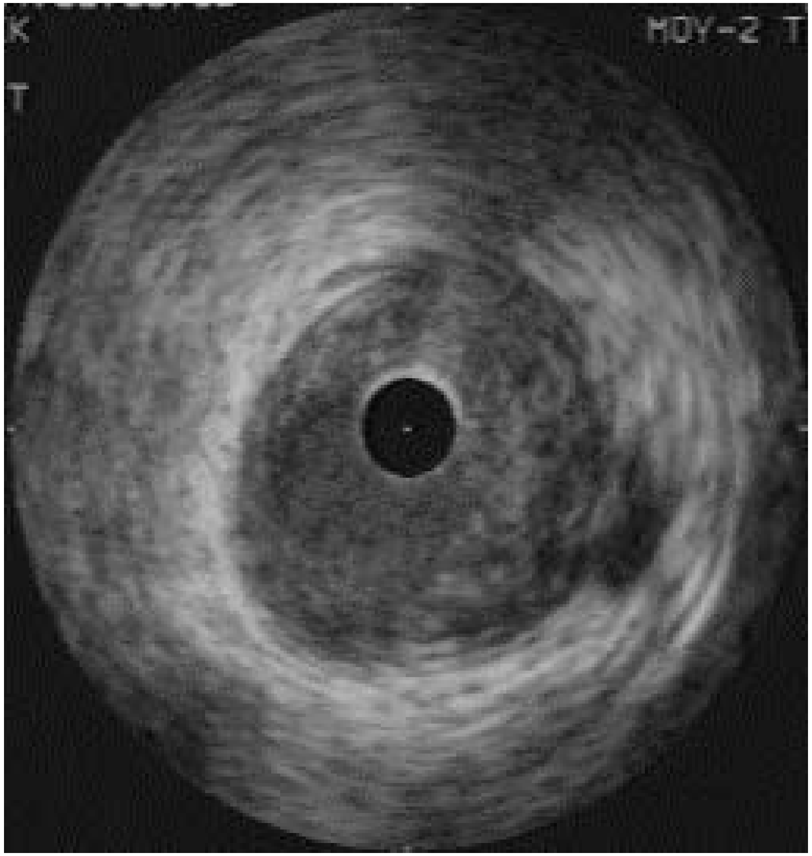
Imaging of the vessels



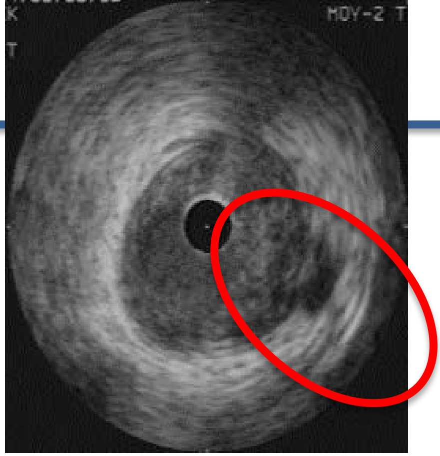
X-ray angiography
skeleton image of artery



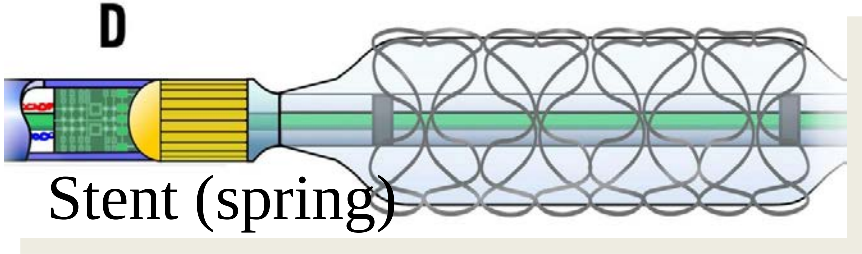
Intravascular Ultrasound
cross-sectional image of artery


DISSECTION

General presentation



- Usefulness in cardiology
  - diagnosis of atherosclerotic coronary artery disease
  - guiding therapeutic procedures (balloon angioplasty, stenting)



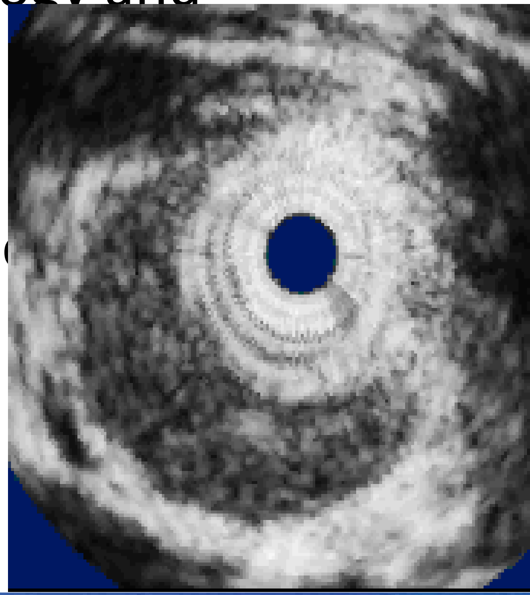
Stent (spring)

- **High resolution** cross-sectional images of vessel walls in **real time**
- Frequency: 30-40 MHz, depth < 10 mm

Main advantages

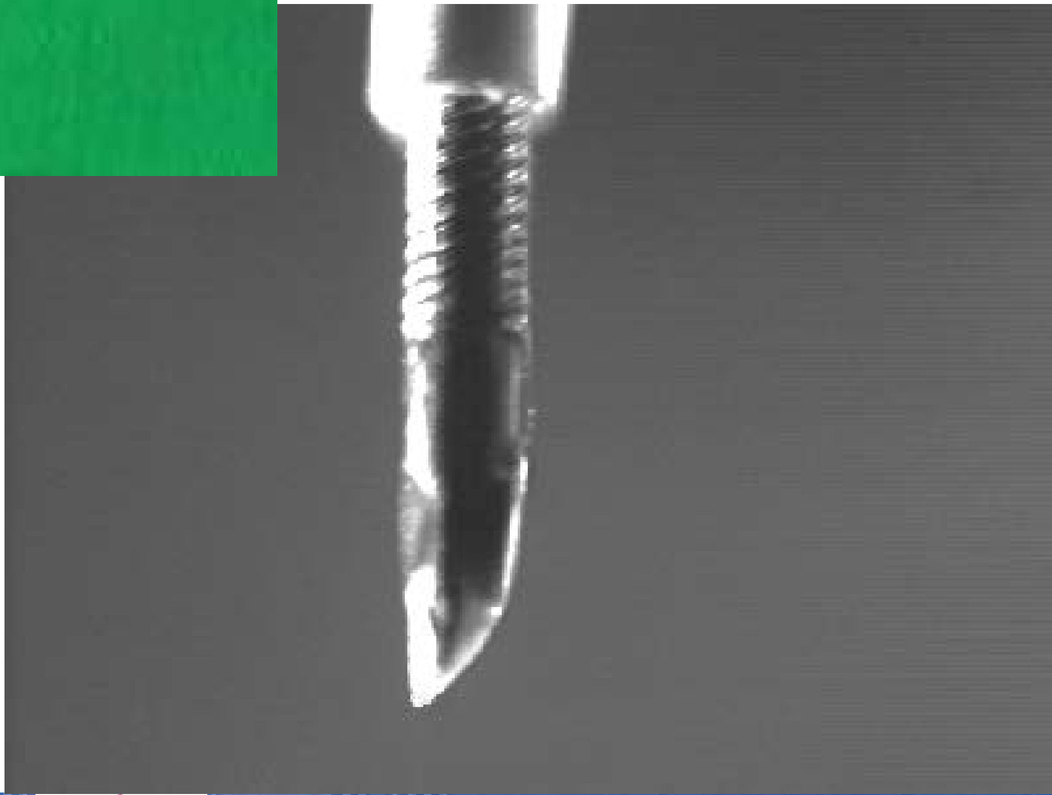
- Time of examination : 5 to 10 mn
- Cost : 800 € (depends on the catheter used)
- Operating room (invasive technique)
- real time cross-sectional images in vivo
- qualitative analysis by characterising roughly plaque components, plaque rupture or acute thrombosis
- depicting atherosclerotic plaque morphology and remodeling

- 2D and 3D quantitative analyses by precis
  - o arterial dimensions (diameter and area)
  - o % stenosis
  - o plaque volume

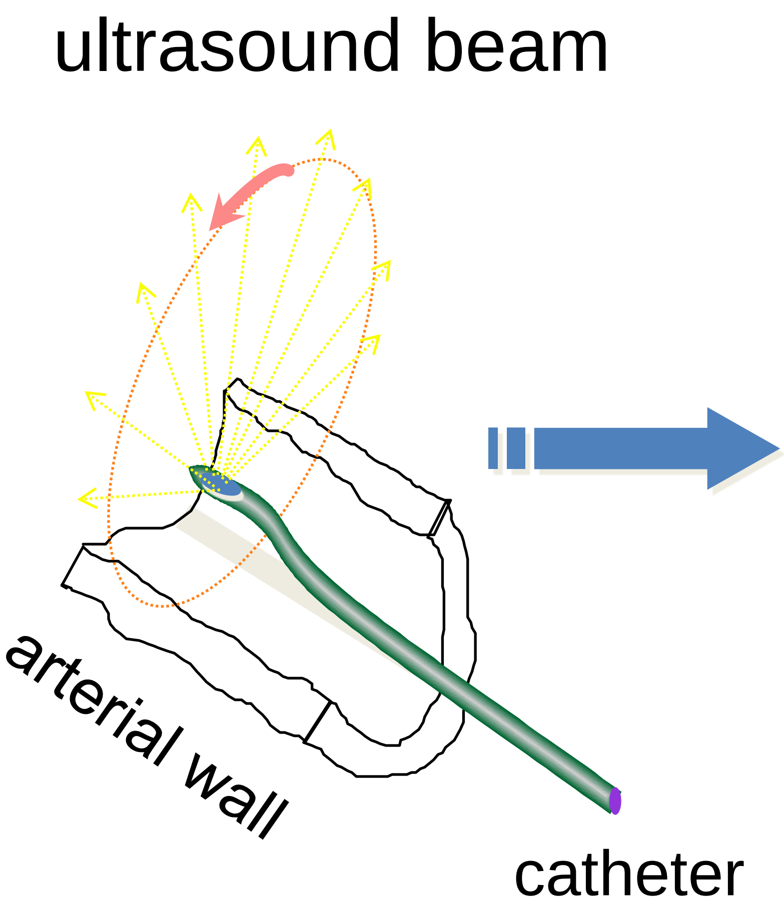


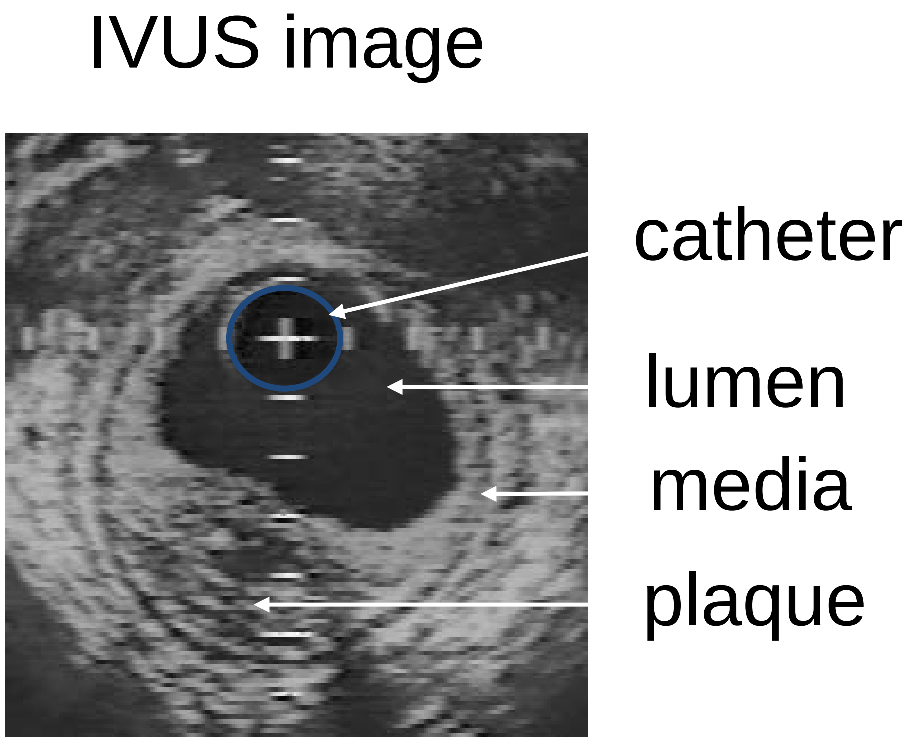
Catheter diameter ≅ 1mm

IVUS is an **invasive** imaging modality



Intravascular Ultrasound (IVUS) Imaging



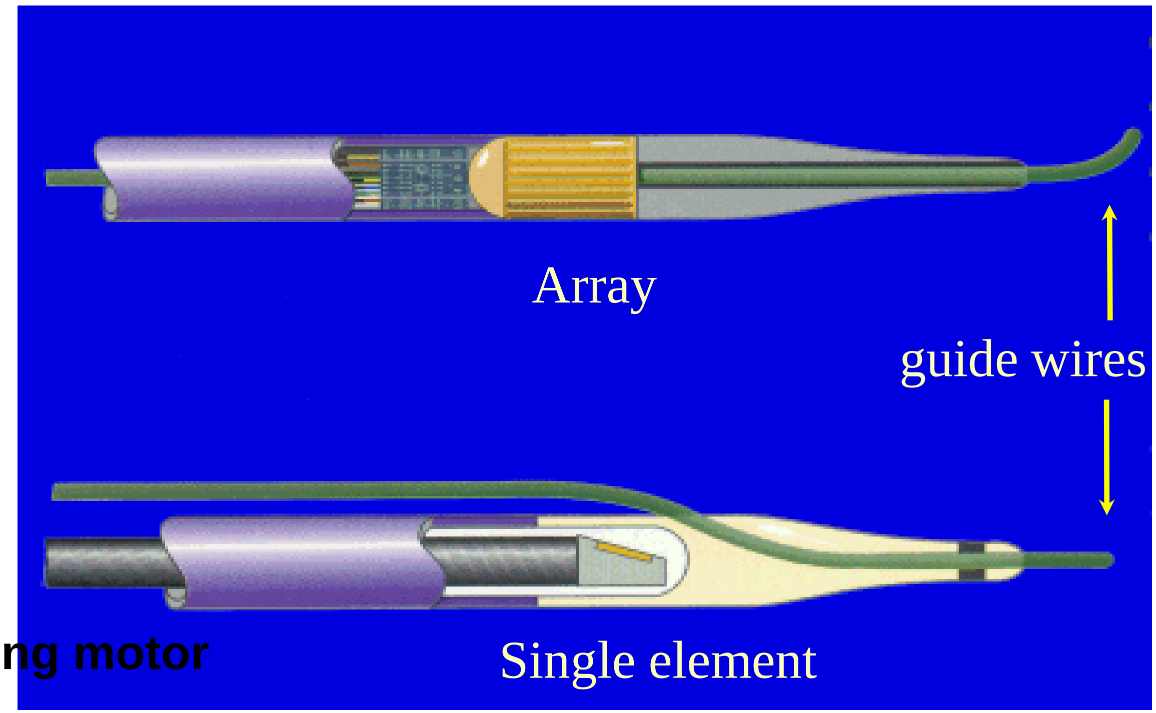


IVUS compare to classic probe

- **Advantages of IVUS**
  - Frequency range: 20 MHz to 40 MHz
    - High frequency: **better resolution** and **more attenuation**
  - Vessel size: some millimeters
  - Wavelength (at 30 MHz): 50 µm
  - Axial resolution (at 30 MHz): 150 µm
  - Lateral resolution (at 30 MHz): 250 µm

- **Drawback**
  - Invasive technique (operating room)

Catheter technologies



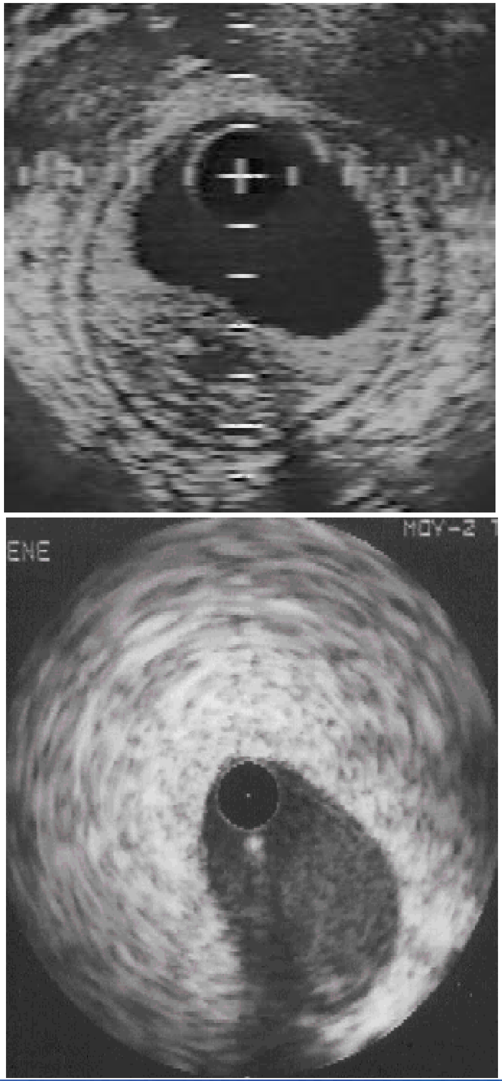
Electronic probe

Mechanical probe

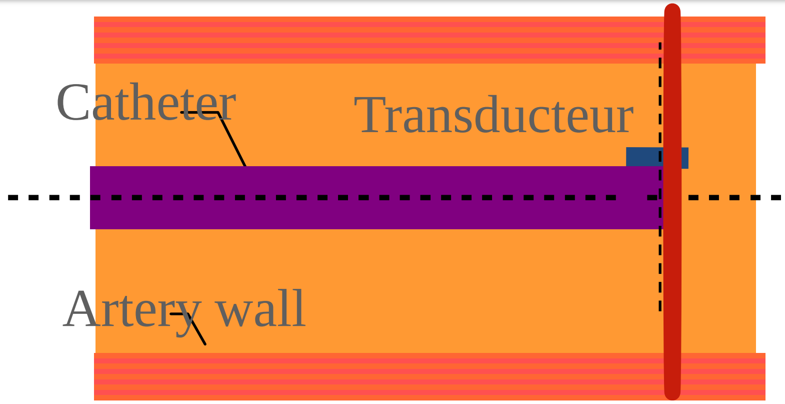
rotating motor

Geometric artefacts

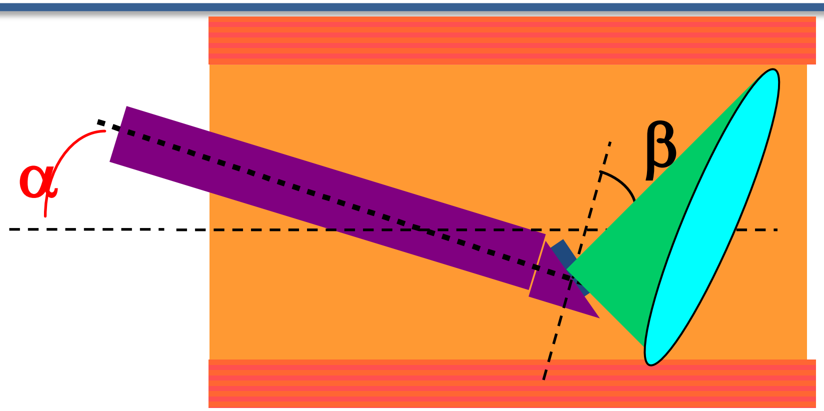
- **Geometric distortion are caused by the position of the ultrasound catheter within the artery**
- **A circular artery is seen on IVUS images as a noncircular vessel represented by more or less complex shapes**



Geometric artefacts



Theoretical case

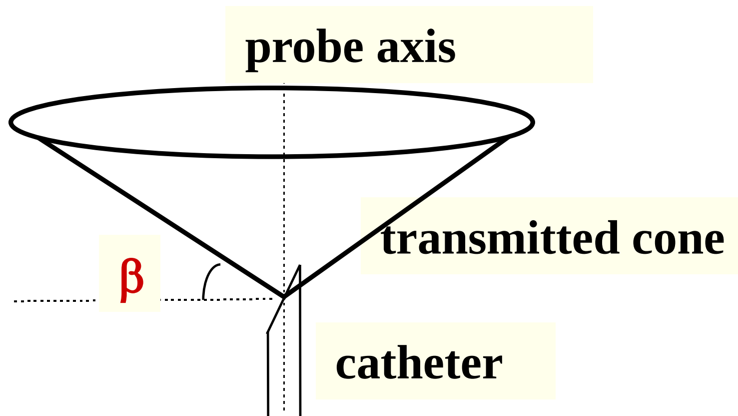


Practical case

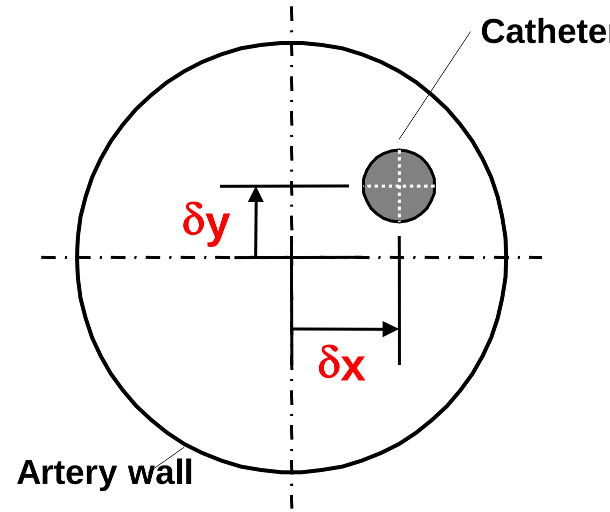
- **The main distortions are due:**
  - to the inclination of the catheter (its long axis is not coaxial the vessel axis), angle α
  - to the off centered position of the catheter (the axis catheter is not located on the axis of the vessel), δx and δy
- **These artefacts are amplified by the geometry of the probe**
  - the origin of the ultrasound beam is not the center of the catheter: ρ
  - the ultrasound beam looks forward (not perpendicular to the long axis of the catheter), angle β

2D artefacts simulation

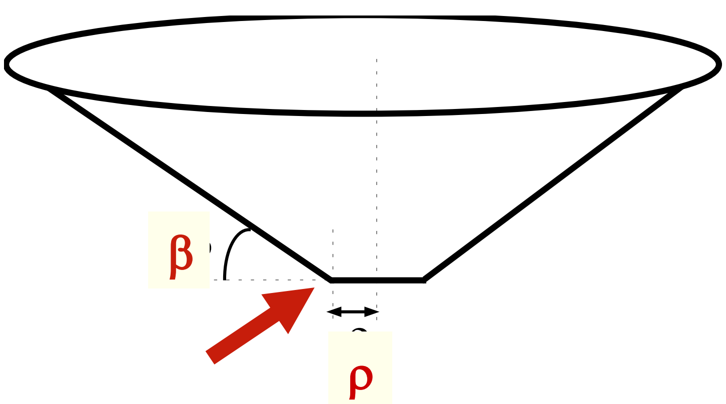
Four origins for geometric artefacts



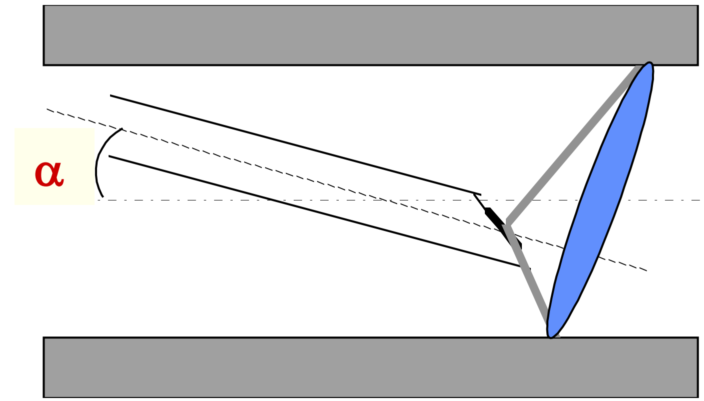
- **Forward inclination**



- **Off centred position**



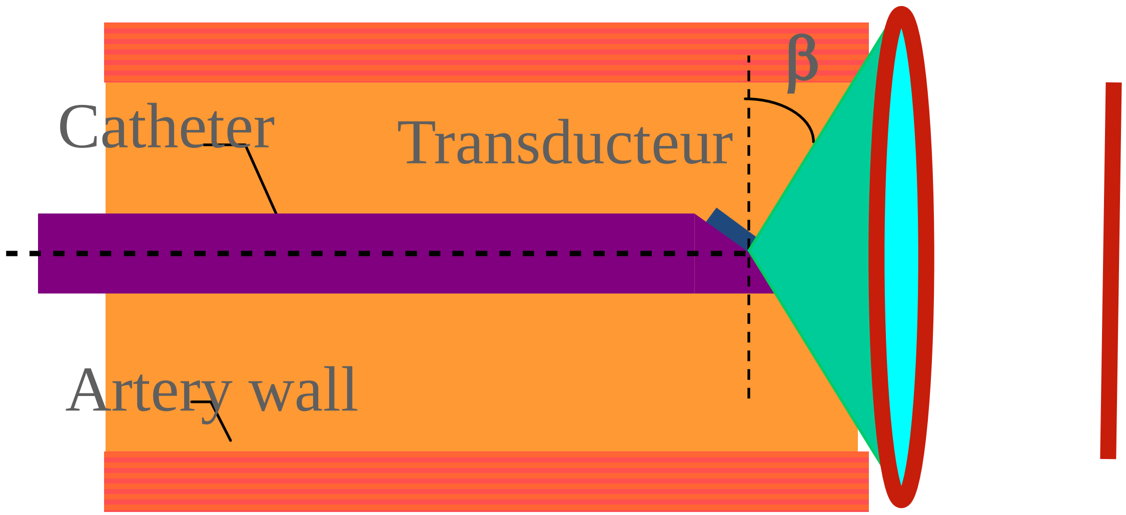
- **Wave transmitted from point out of center**

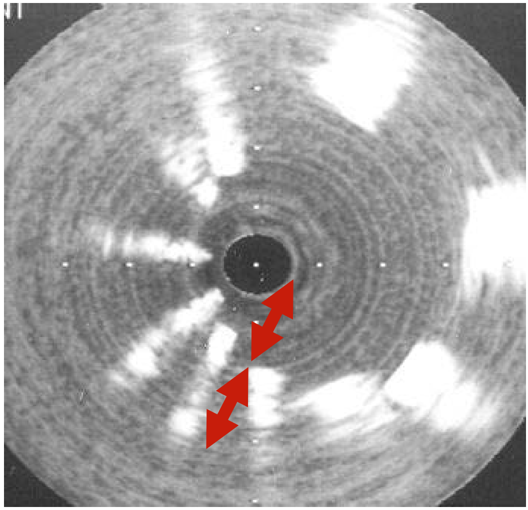


- **Inclination of the catheter**

Forward inclination angle β

Transmitting cone swept by the ultrasound beam





$\beta = 15^\circ$ $\beta = 0$

Multiple reflexions

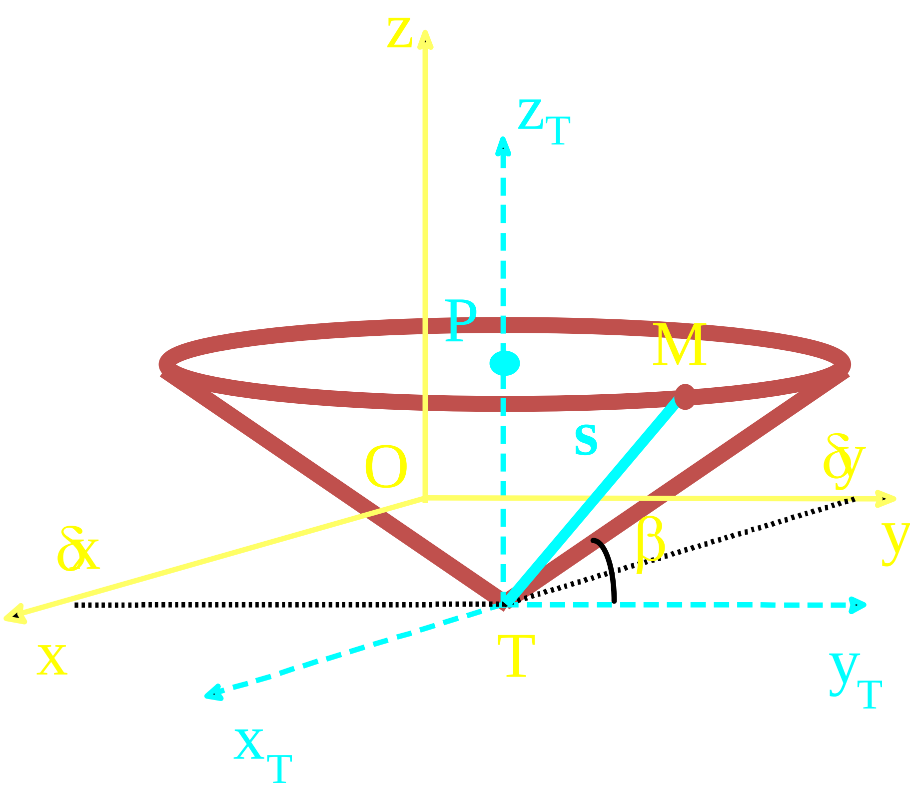
- **The forward inclination of the piezoelectric element avoids direct reflection on the vessel wall and the multiple reflection. The scattering is reinforced.**

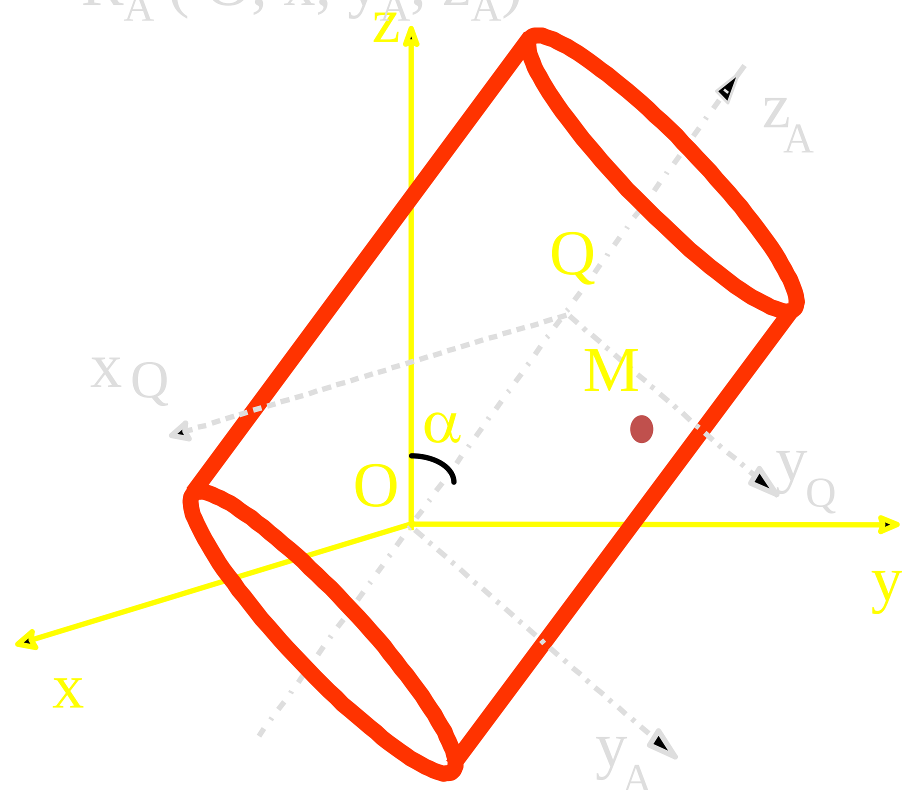
Modelisation of the geometry

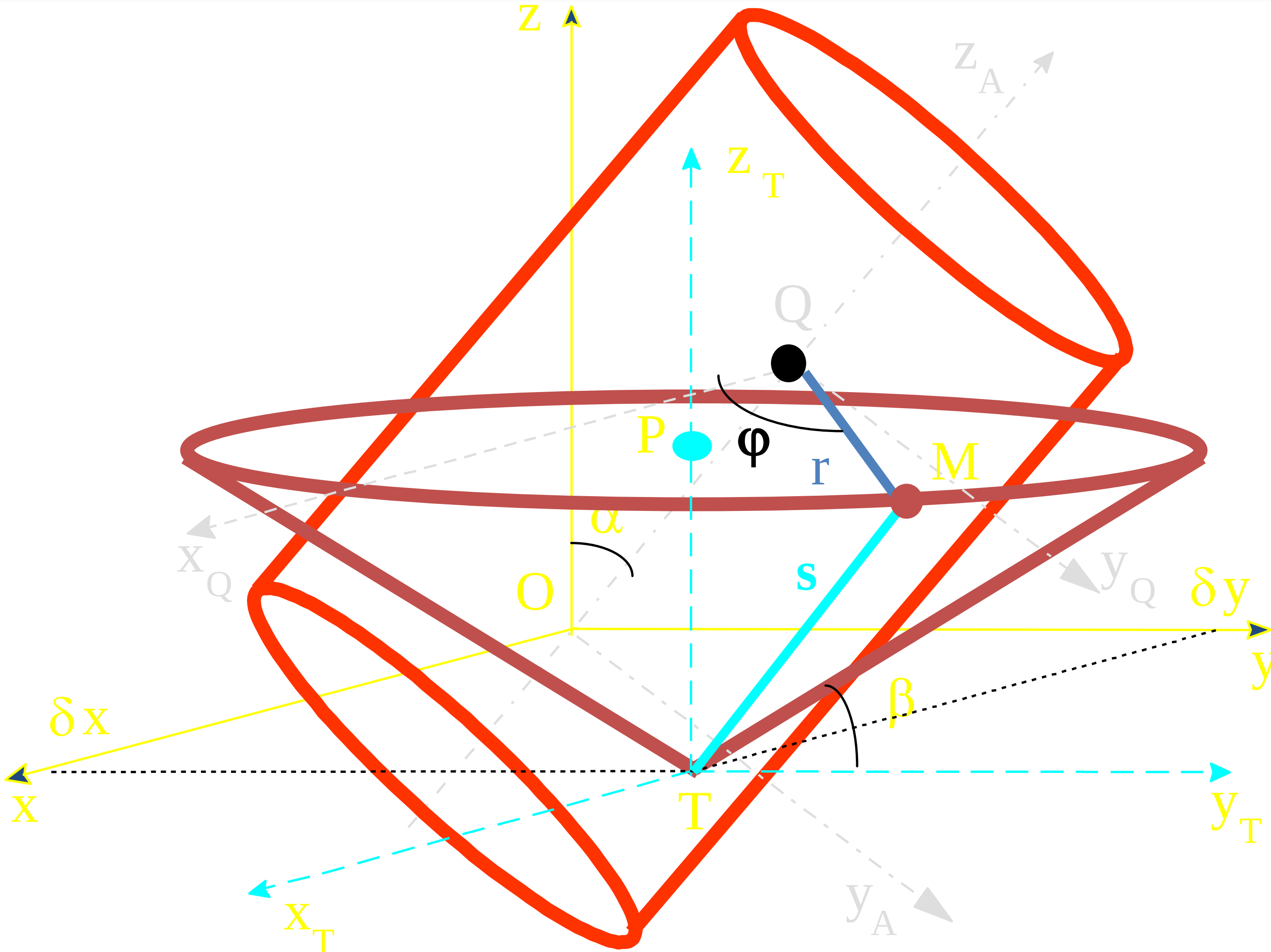
Reference coordinate system: R (O, x, y, z)

Probe coordinate système:
R_T (T, x_T, y_T, z_T)

Artery coordinate systeme:
R_A (O, x, y_A, z_A)

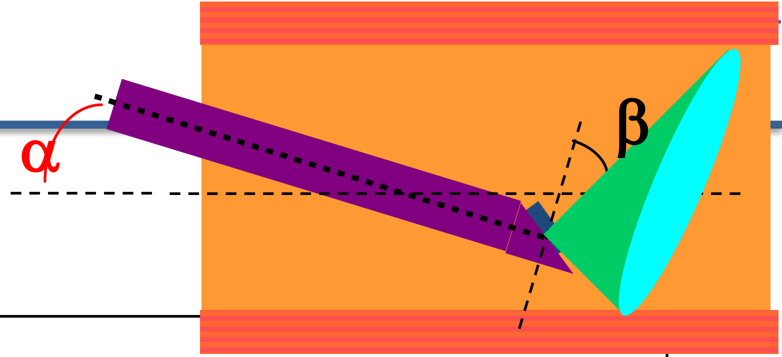


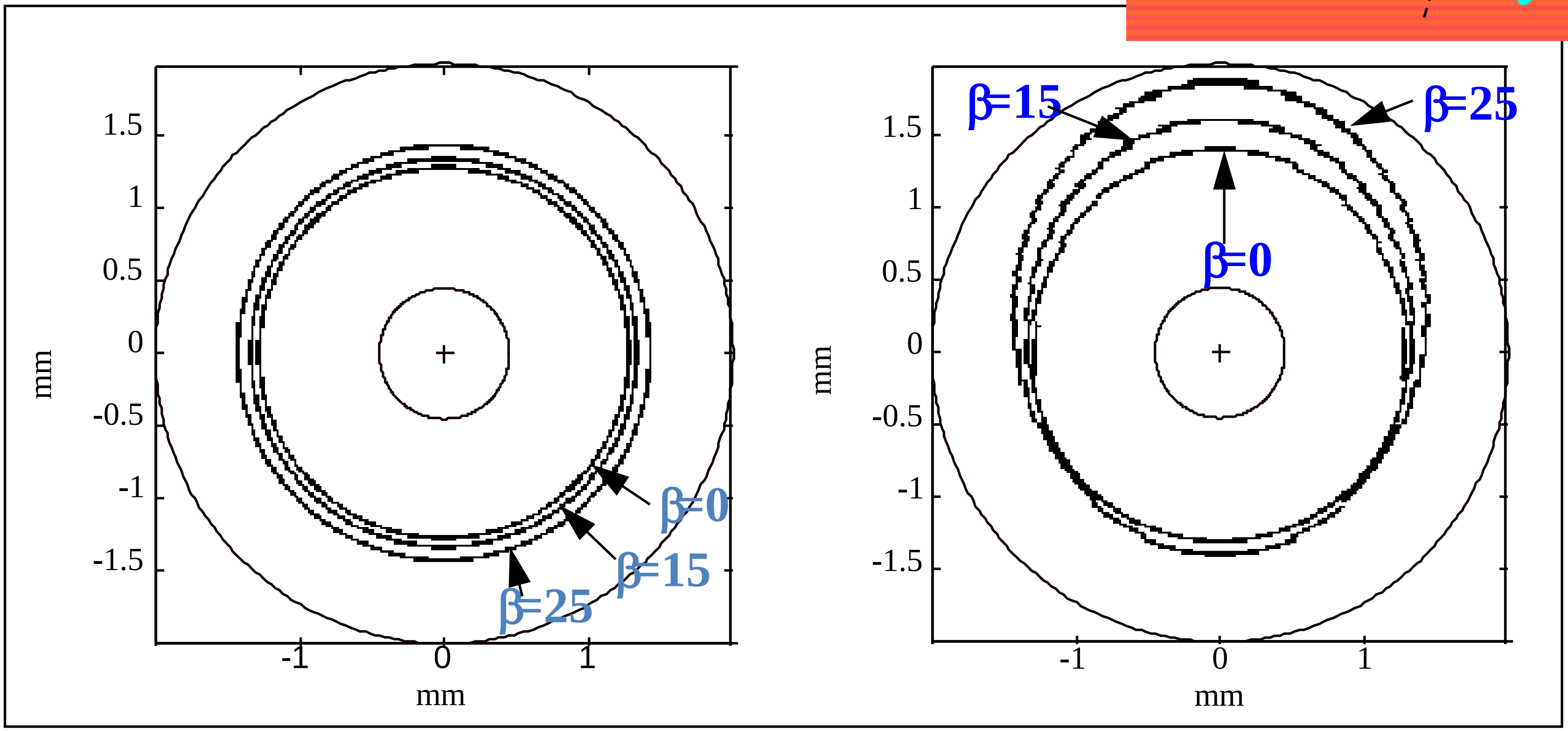




Delachartre P., Cachard C. and al., Ultrasound in Medicine and Biology, 1999, vol. 25

Forward inclination angle β



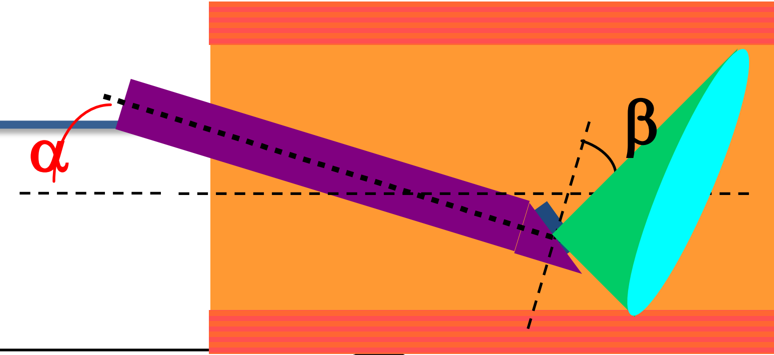


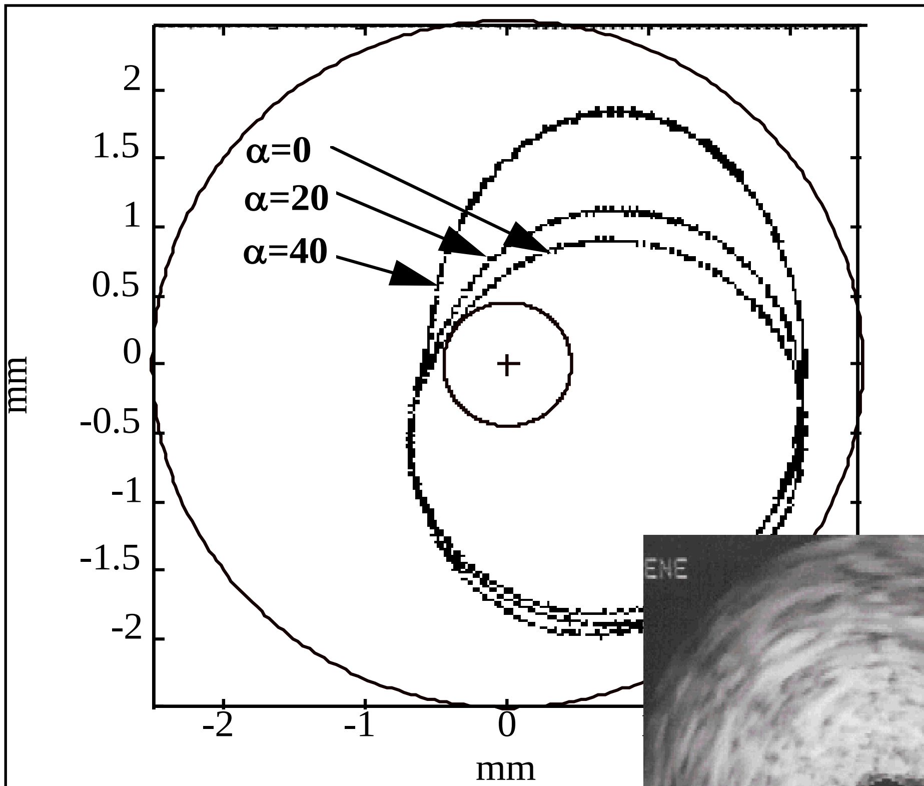
inclination $\alpha = 0$ inclination $\alpha = 20^\circ$

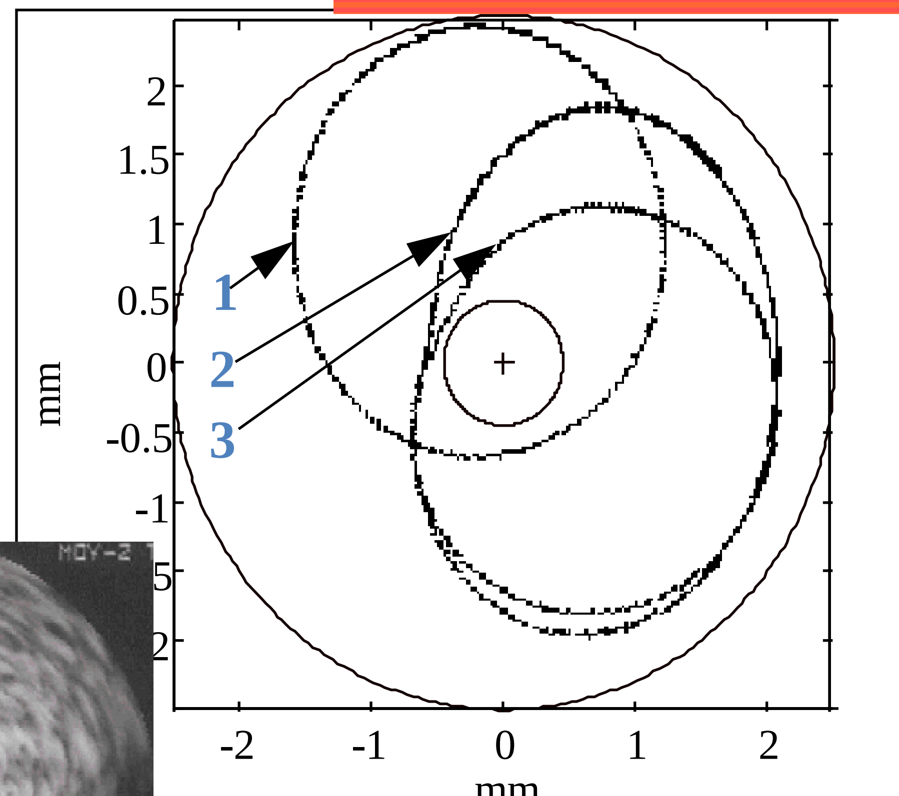
off centered position $\delta x = 0$ mm, $\delta y = 0$ mm

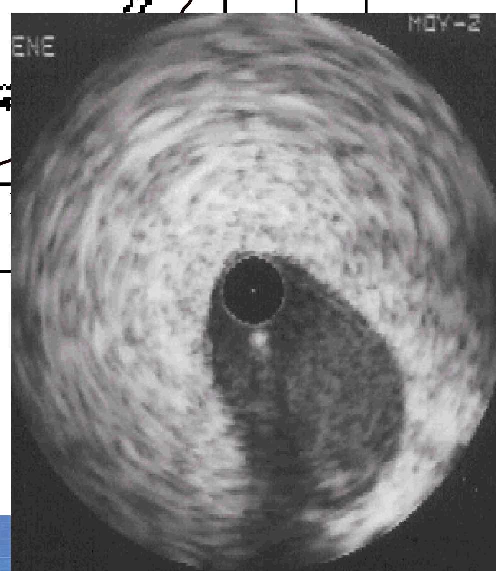
off axis of transmition point $\rho = 0.4$ mm

Inclination of the catheter axis: α





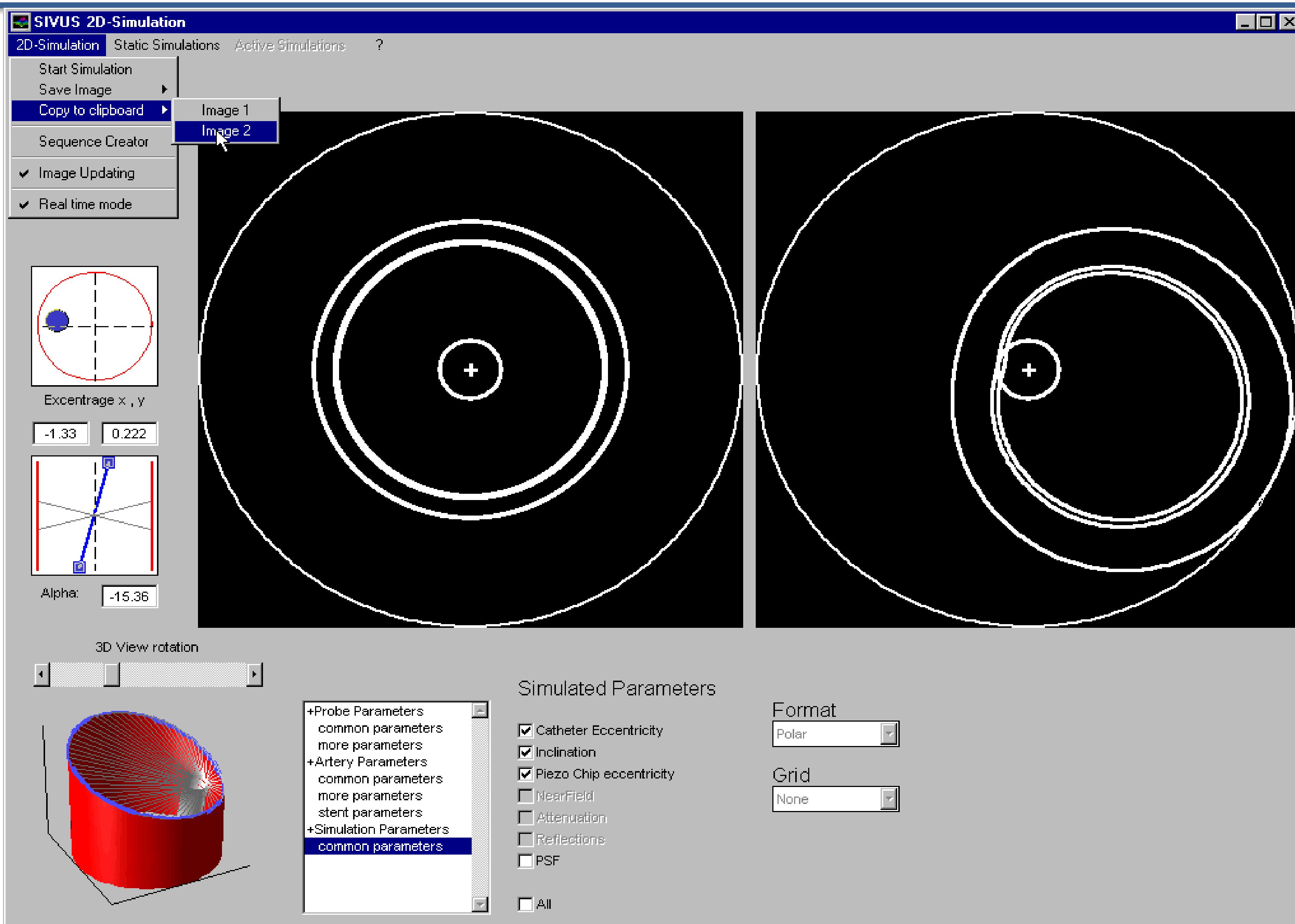




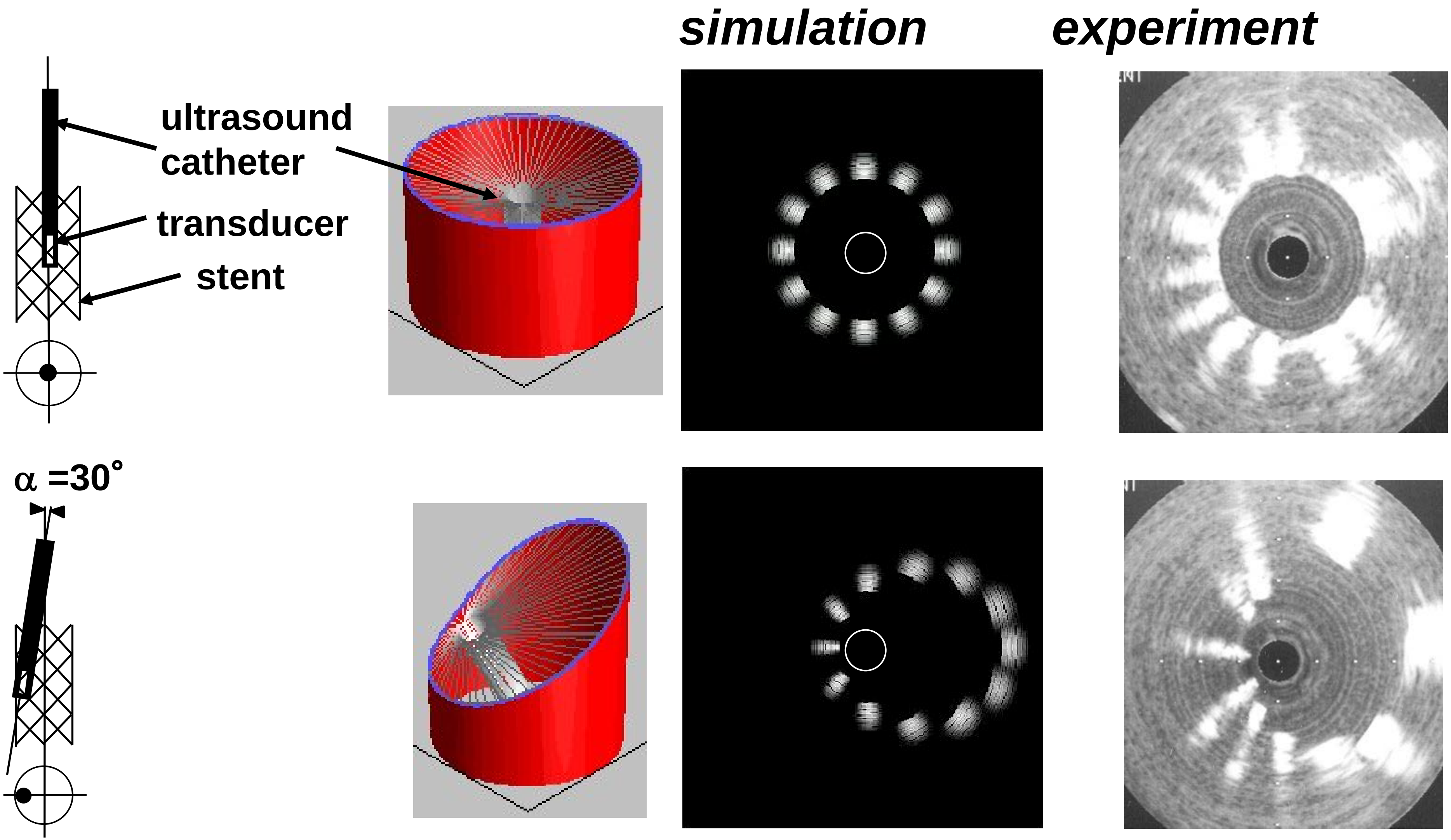
δx = - 0.7 mm, δy = 0.5 mm
β = 15°
ρ = 0.4 mm

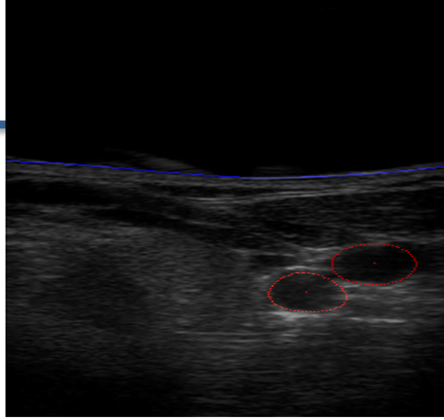
1 : α = 20° , δx = 0.2 et δy = - 0.7;
2 : α = 40° , δx = - 0.7 et δy = 0.5;
3 : α = 20° , δx = - 0.7 et δy = 0.5
(β=15° et ρ = 0.4 mm).

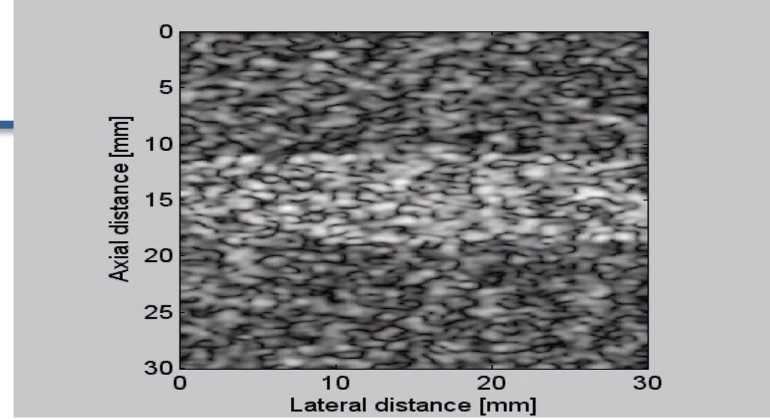
2D simulation: screen capture



Geometric artifacts







Nonlinear ultrasound imaging and multiplulses imaging modalities

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The advances in multipulses and nonlinear ultrasound imaging modalities

- **Advanced in** (Nonlinear) **Ultrasound**
  - Ultrasound Contrast Agent
  - Nonlinear ultrasound
- **Improvement of ultrasound imaging or Nonlinear imaging**
  - Harmonic imaging
  - Multipulses imaging or Contrast (nonlinear) imaging
  - Generalization of multi-pulse techniques
  - Influence of scatterer motion to phased multipulses method
- **CREANUIS: Simulation of nonlinear ultrasound images**

Nonlinear propagation

Nonlinear propagation

Creatis

The motion equation

$$\rho \frac{\partial \vec{u}}{\partial t} + \nabla p = 0$$

The pressure is expanded using the Taylor series

$$p = p_0 + \frac{A}{1!}\left(\frac{\rho-\rho_0}{\rho_0}\right) + \frac{B}{2!}\left(\frac{\rho-\rho_0}{\rho_0}\right)^2 + \frac{C}{3!}\left(\frac{\rho-\rho_0}{\rho_0}\right)^3 + \dots$$

linear

The celerity

$$c = c_0\left(1+\frac{B}{2A}\frac{u}{c_0}\right)^{\frac{2A}{B}+1} \approx c_0 + \left(1+\frac{B}{2A}\right)u = c_0 + \beta u \quad \Rightarrow \quad \beta = 1 + \frac{B}{2A}$$

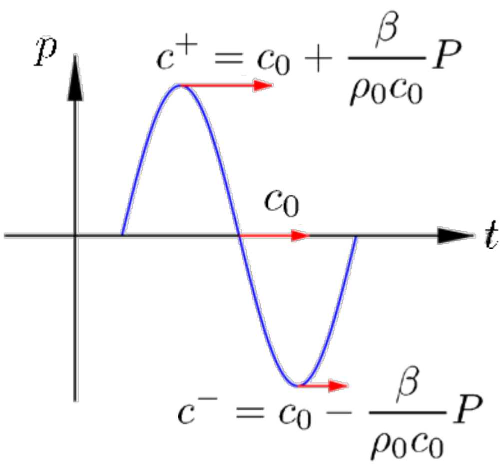
Nonlinear coefficient

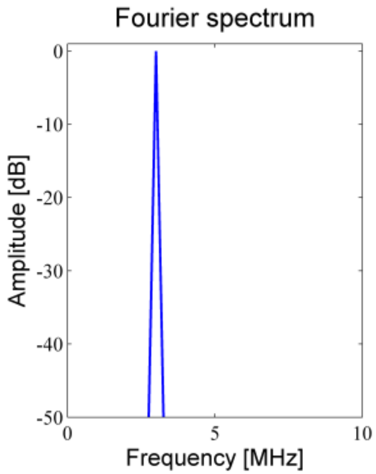
Nonlinear parameter

Nonlinear propagation: celerity shift

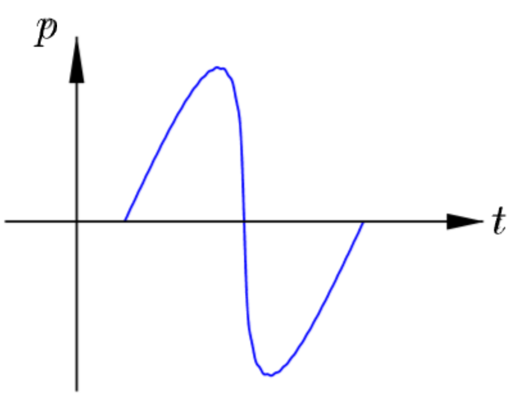
$$c = c_0\left(1+\frac{B}{2A}\frac{u}{c_0}\right)^{\frac{2A}{B}+1} \approx c_0 + \left(1+\frac{B}{2A}\right)u = c_0+\beta u \Rightarrow \beta = 1+\frac{B}{2A}$$

Nonlinear coefficient — Nonlinear parameter

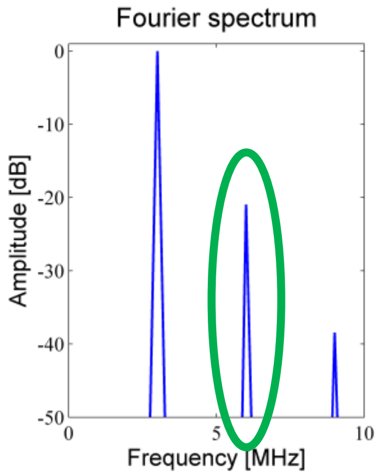




f_0

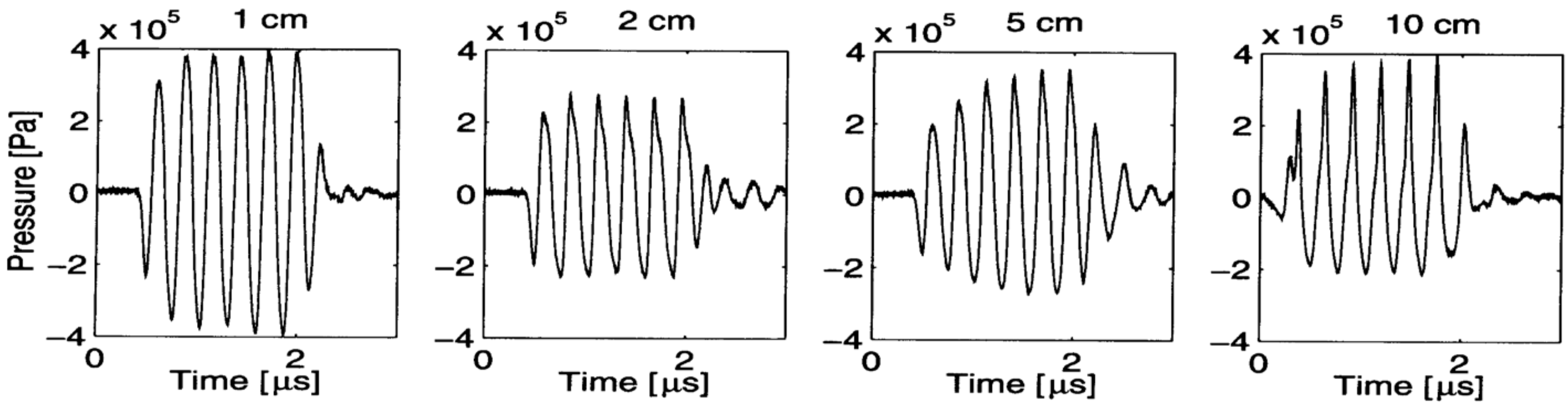


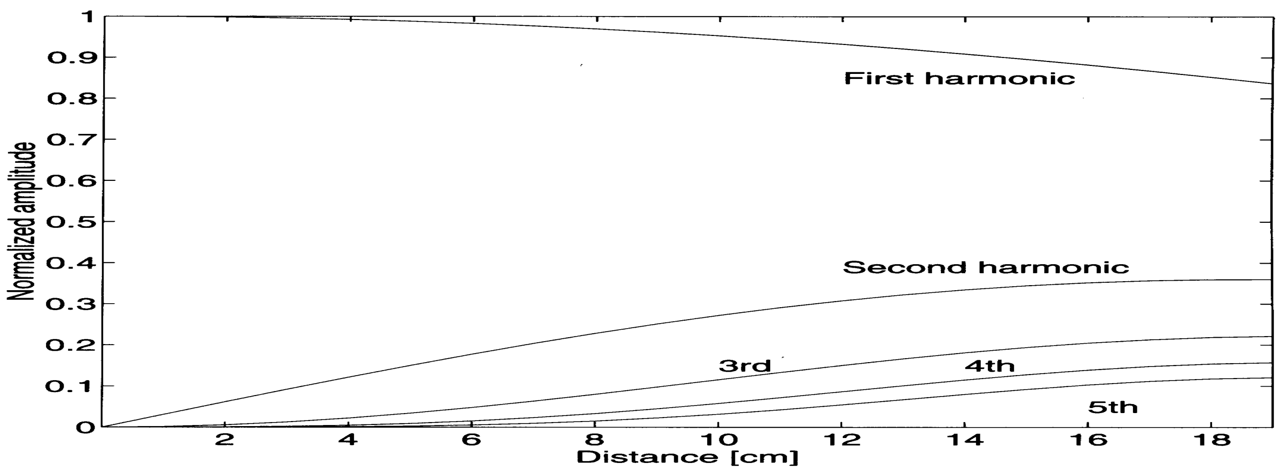
After propagation



$f_0 + 2 f_0 + 3 f_0 + \ldots$

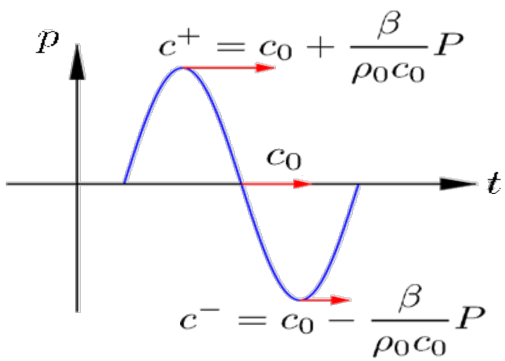
Non linear propagation

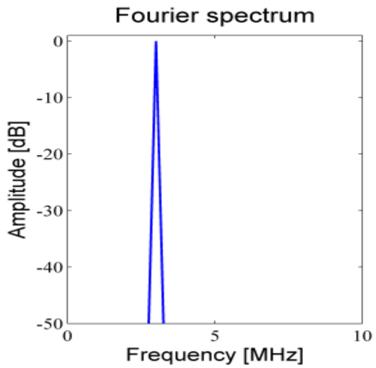




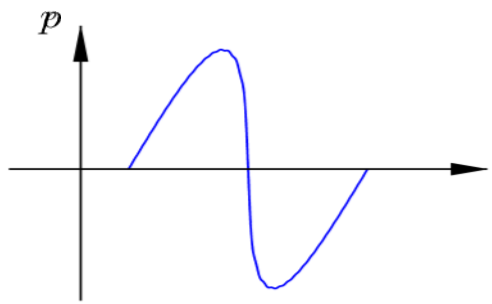
Nonlinearity: tissue and contrast agent

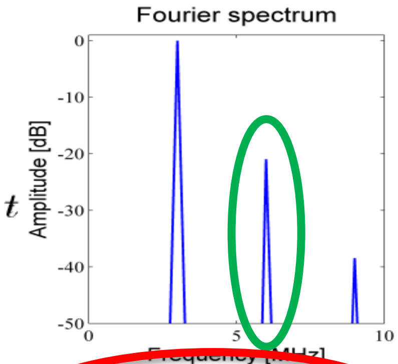
- **Tissue**





f_0

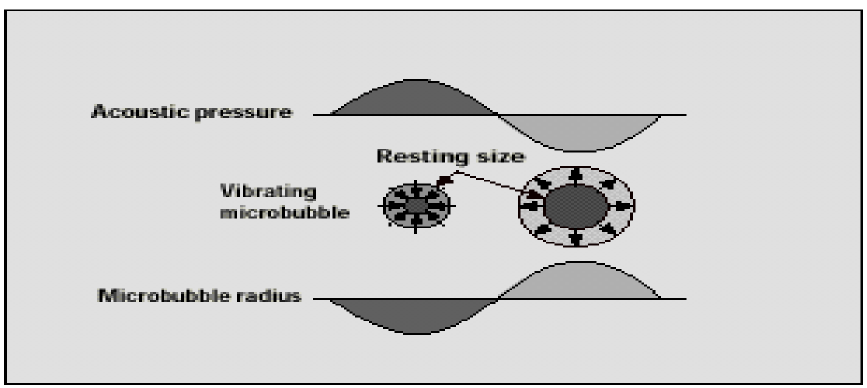


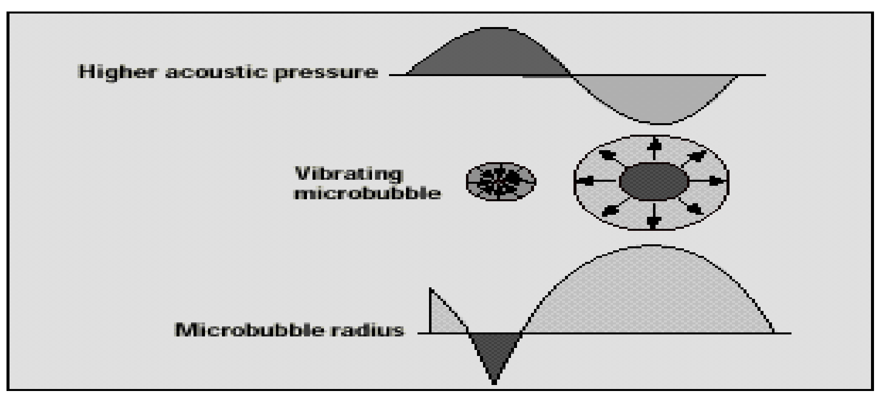


After propagation

$f_0 + 2\ f_0 + 3\ f_0 + ...$

- **Contrast agent**





Linear scattering: the vibration is symmetric

Non linear scattering: the vibration is asymmetric

f_0

$f_0 + 2\ f_0 + 3\ f_0 + ...$

Creatis

It is a signal processing problem:

- **higher SNR,**
- **selection of frequency band: f_0 , 2 f_0, 3 f_0...**

Which part of the backscattered signal ?

higher $\frac{\text{signal}}{\text{noise}}$

Which part of the backscattered signal ?

Specific imaging

higher $\frac{\text{signal}}{\text{noise}}$

Which part of the backscattered signal ?

Which part of the backscattered signal ?

One scan

Two or more scans

Transmitted pulses and signal processing

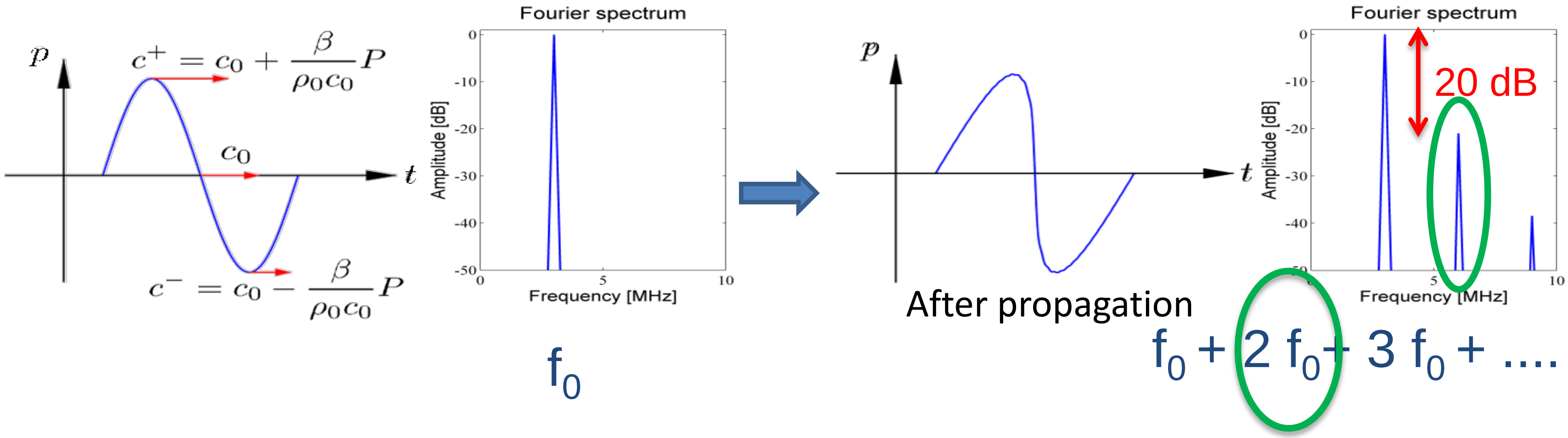
Harmonic Imaging (without UCA)

- Harmonic or Tissue Harmonic Imaging

  (without Ultrasound Contrast Agent)

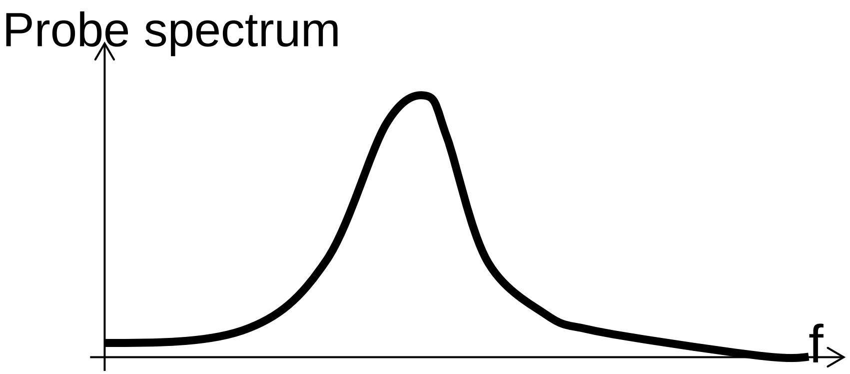
  ✓ Imaging of the second harmonic backscattered to the probe (non-linear propagation in tissue)

- **Tissue**



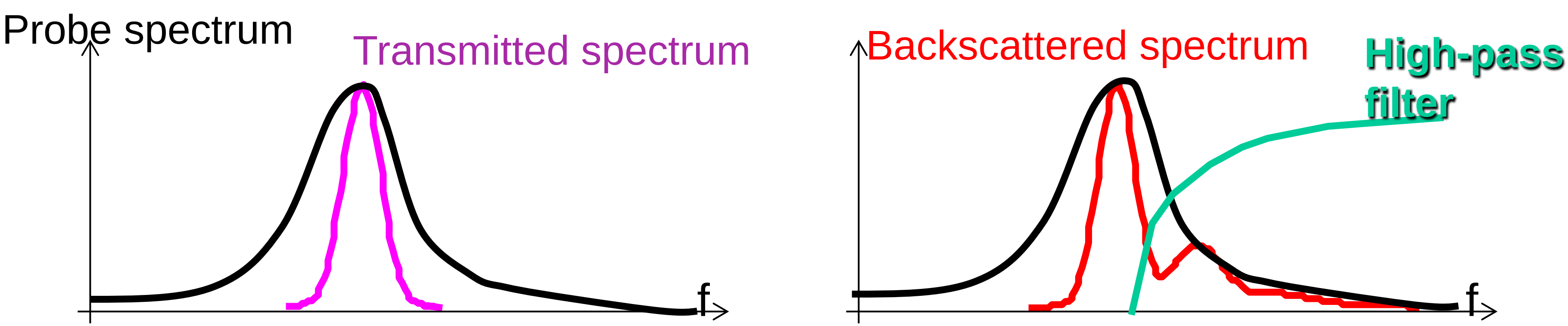
- A high pass filter at 2 f_0

How to image the second harmonic?



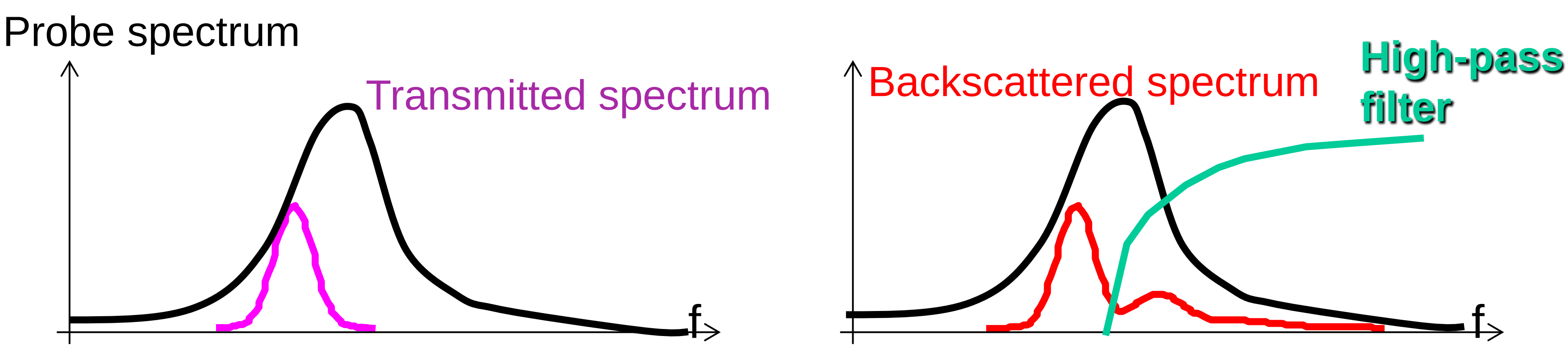
- An ultrasound probe is a passband filter

➢Narrowband transmitted signal

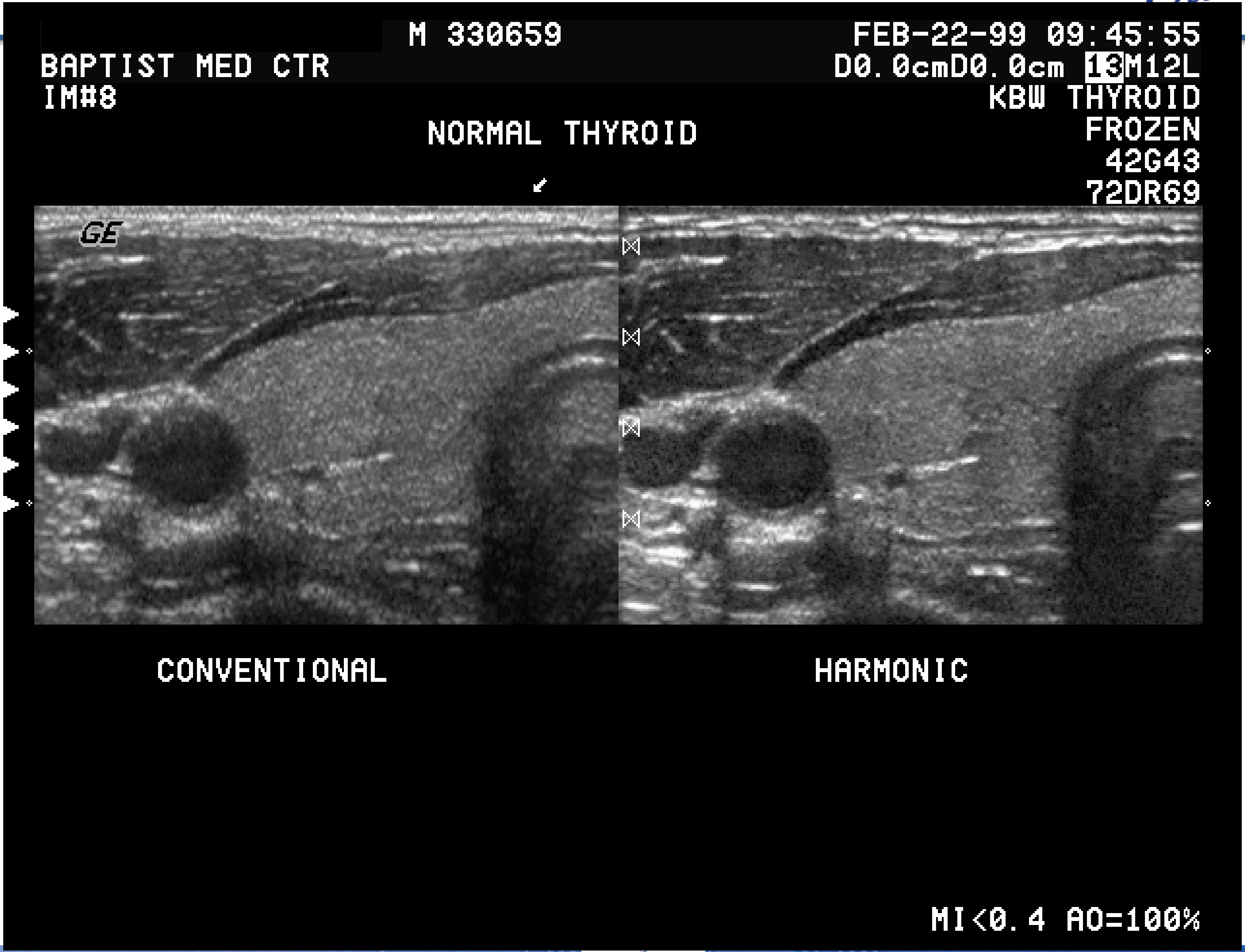


Resolution is decrease

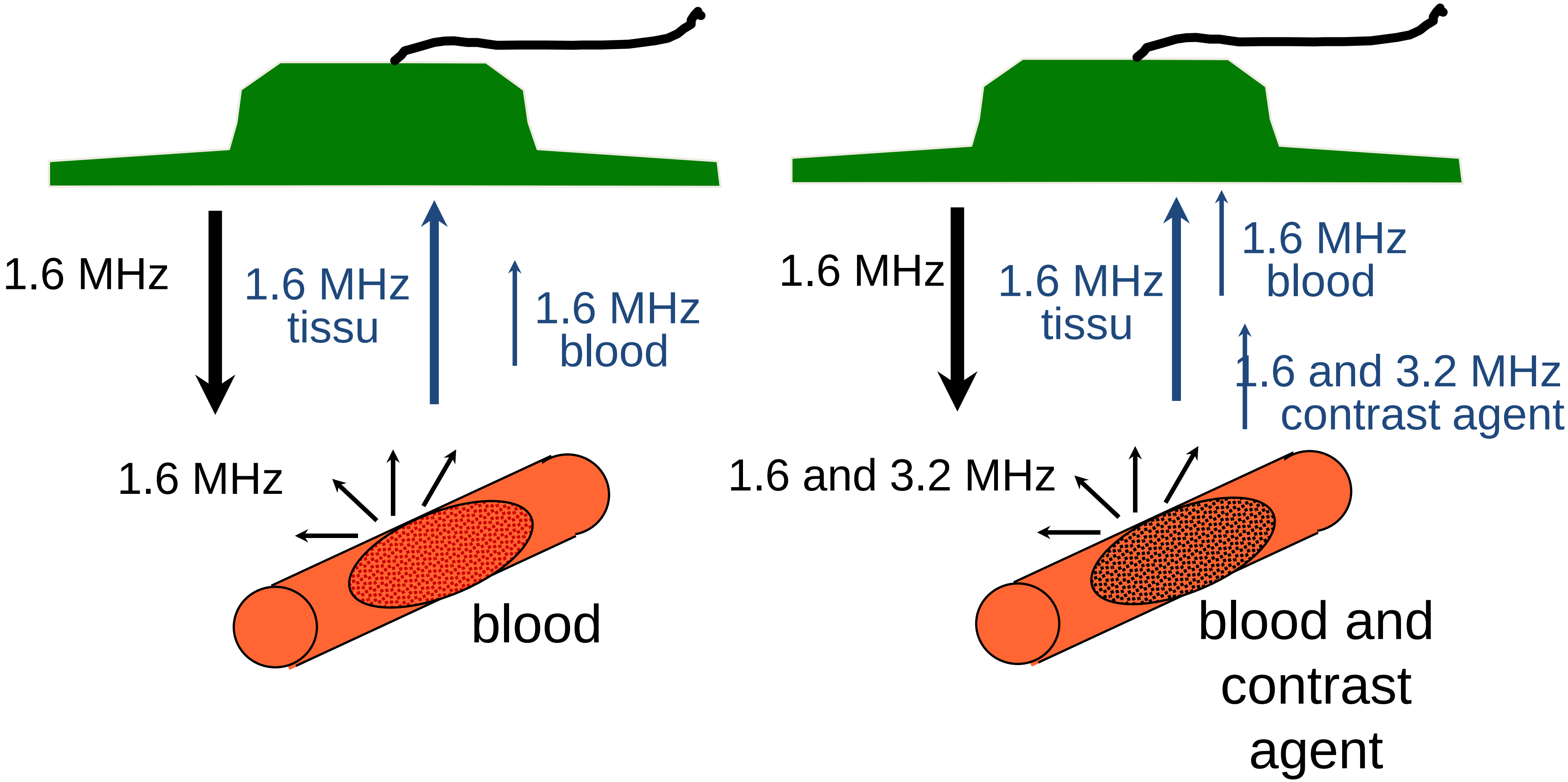
➢Transmitted signal in low part of transducer bandpass



Dynamic is less


M 330659
FEB-22-99 09:45:55
BAPTIST MED CTR
D0.0cmD0.0cm 13M12L
IM#8
KBW THYROID
NORMAL THYROID
FROZEN
42G43
72DR69
GE
CONVENTIONAL
HARMONIC
MI<0.4 AO=100%

Harmonic imaging with contrast agent

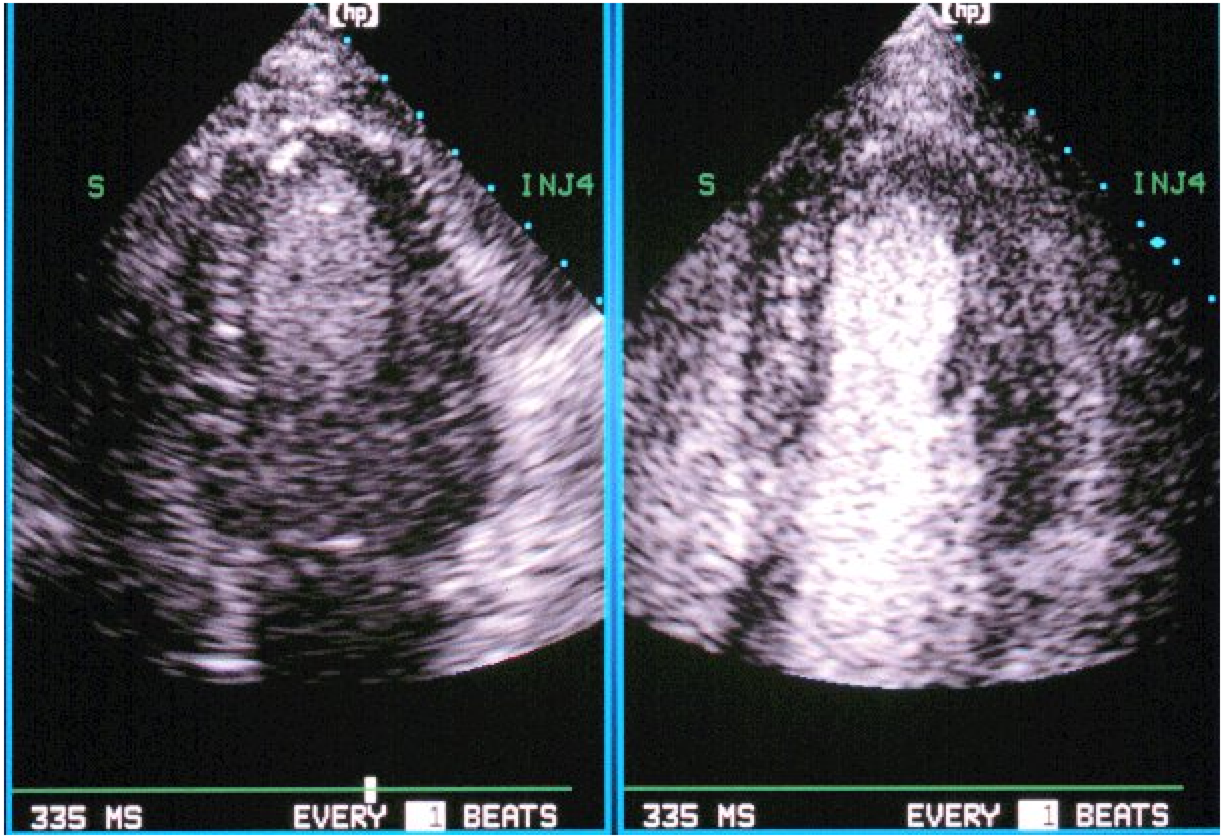


Assumption

Non linearity is higher in contrast agent than in tissue

Non linearity due to propagation is negligible

Harmonic imaging



fundamental second harmonic

Outline

- **Basic of** (Linear) **ultrasound and medical ultrasound imaging**
- **Advanced in** (Nonlinear) **Ultrasound**
  - Ultrasound Contrast Agent
  - Nonlinear ultrasound
- **Improvement of ultrasound imaging or Nonlinear imaging**
  - Harmonic imaging
  - Multipulses imaging or Contrast (nonlinear) imaging
    - Pulse inversion (PI)
    - Second Harmonic Inversion (SHI)
  - Generalization of multi-pulse techniques
  - Influence of scatterer motion to phased multipulses method
- **CREANUIS: Simulation of nonlinear ultrasound images**

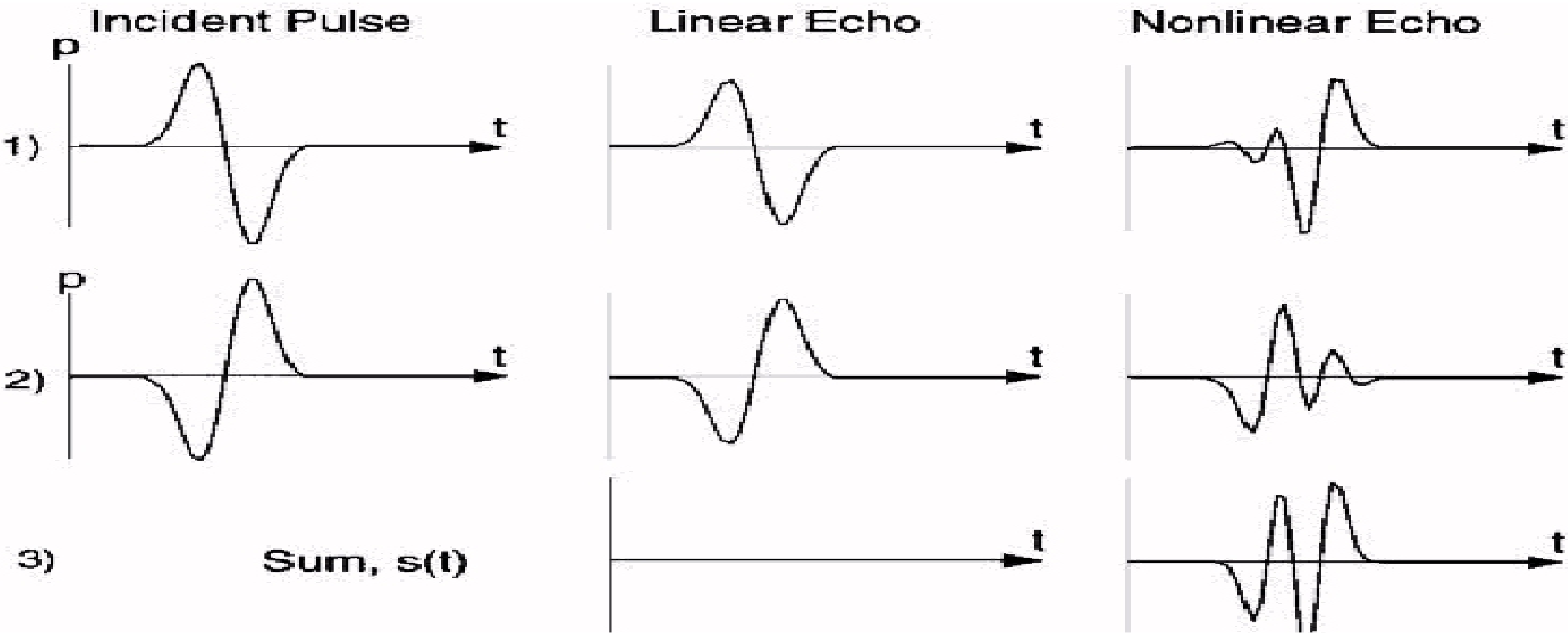
Multi-pulse techniques

Multi-pulse techniques (with ultrasound contrast):

- Amplitude modulation (AM), Brock-Fisher *et al,* 1996
- Pulse inversion (PI), Simpson *et al.,* 1999
- Phase coded sequences (PCS), Wilkening *et al.,* 2001
- Contrast pulse sequence (CPS), Phillips, 2001
- Pulse inversion amplitude modulation (PIAM), Eckersley *et al.,* 2005
- Second harmonic inversion (SHI), Pasovic *et al.,* 2011

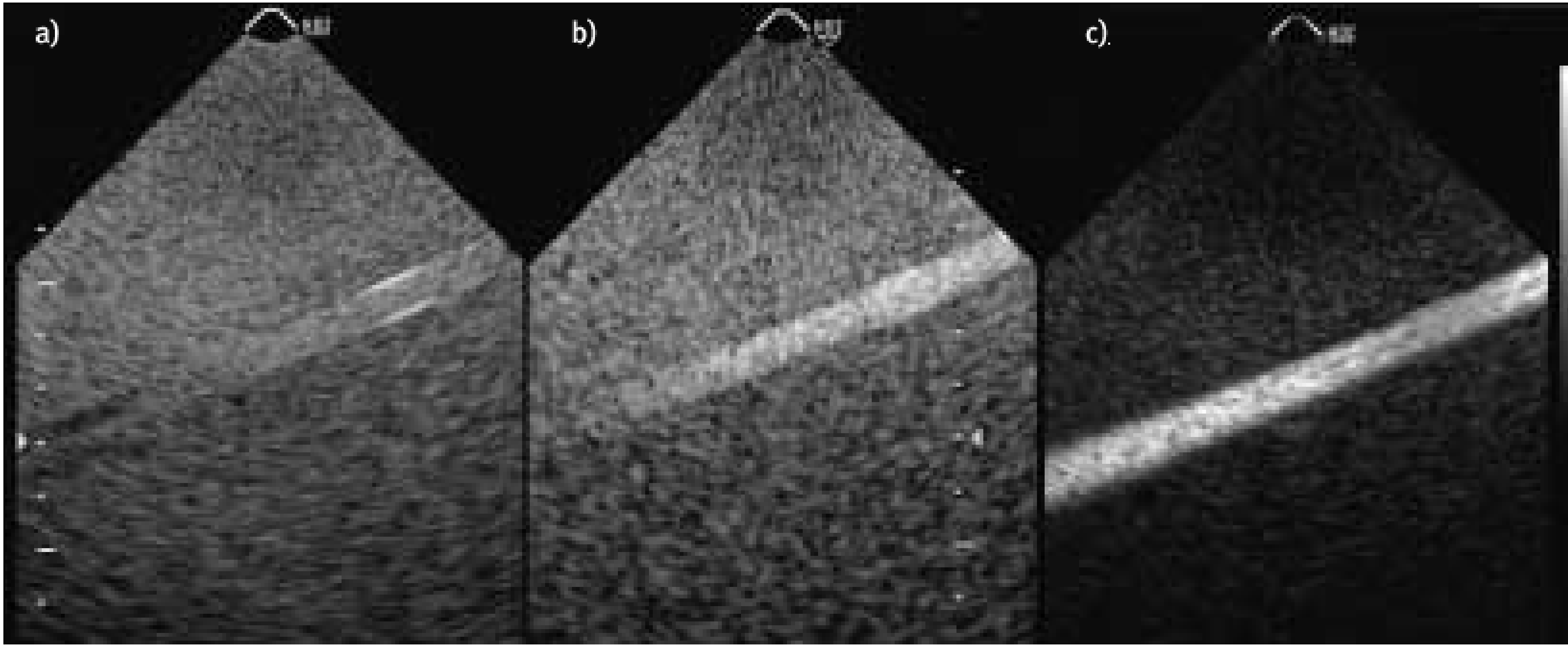
Pulse inversion imaging

- 2 incident pulses whith a phase shift $\varphi = 180^\circ$



Simpson, IEEE TUFFC, vol. 46, no. 2, march 1999

Pulse inversion imaging



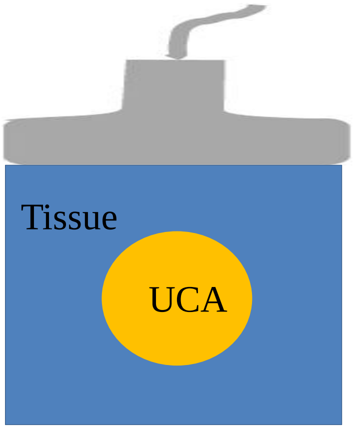
Conventional US imaging

Harmonic Doppler imaging

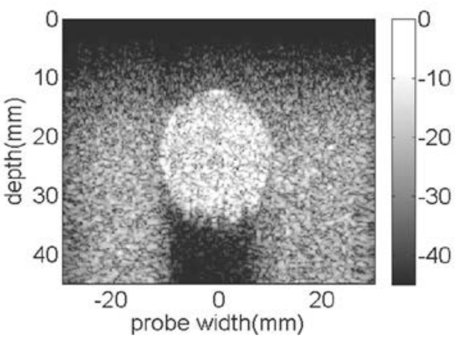
Pulse inversion Doppler imaging

With the courtesy of Piero Tortoli (Firenze University, Italy)

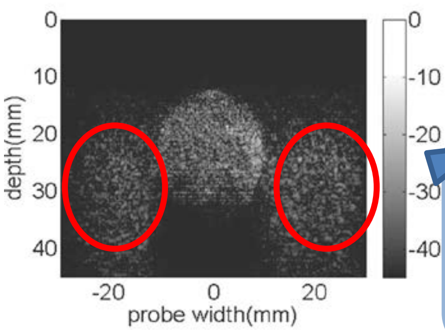
SHI: Second Harmonic Inversion



B Mode Image



Second Harmonic Image



- Tissue was previously regarded as linear (only f_0 signal exists)
- However, tissue also generates harmonic signals during the propagation: $f_0+2f_0+\dots$
- Discrimination between UCA and tissue is reduced.

Undesired effect

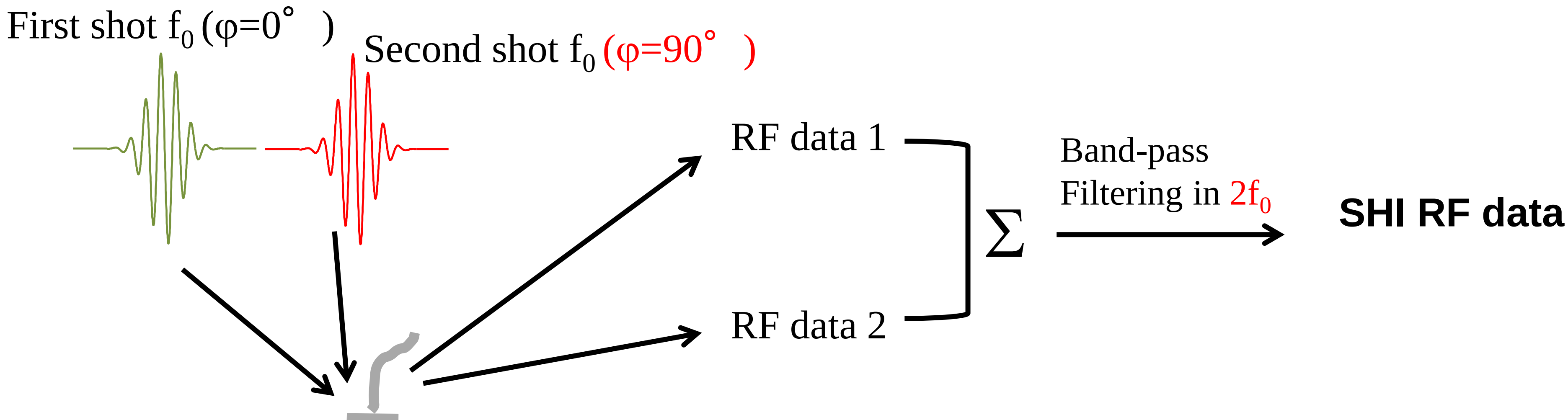
SHI objectives:

- ❖ **Decrease the second harmonic generated in tissue**
- ❖ Increase the discrimination between UCA and tissue (CTR)

SHI: Second Harmonic Inversion

Method

- As PI, 2 incident pulses but the phase shift is φ = 90°



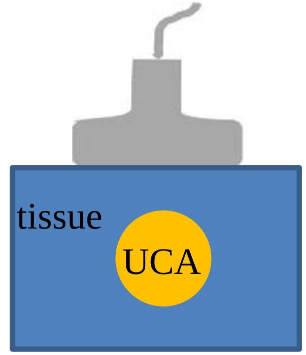
$$RF_{sum} = RF_1(\varphi = 0) + RF_2(\varphi = \frac{\pi}{2})$$

Pasovic, *Physics in Medicine and Biology*, 2011

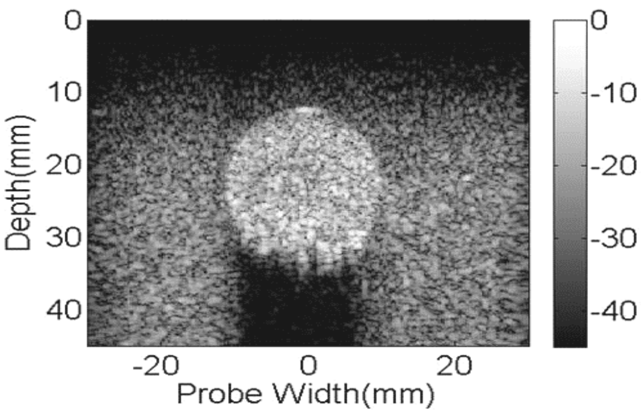
SHI: Second Harmonic Inversion

Experimental evaluation of SHI

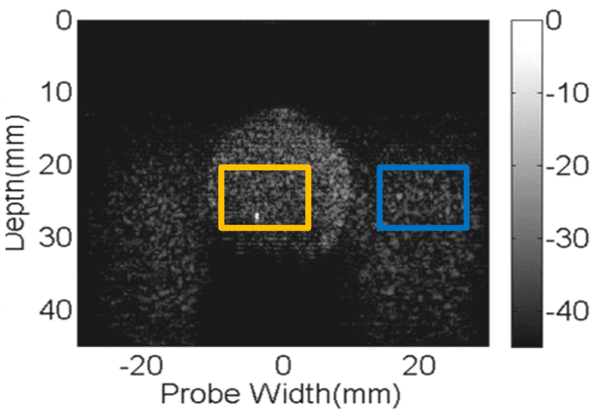
- A tissue mimicking phantom (5% agar; 1% silice)
- UCA (SHU 508A)
- Transmitted frequency: 5 MHz



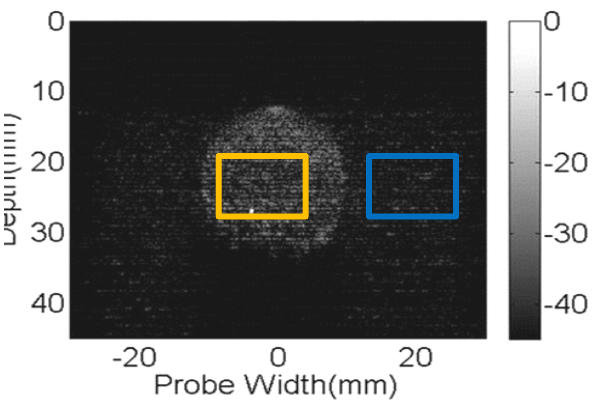
B mode



Second harmonic



SHI



$$CTR = 20\log\frac{P_2^{UCA}}{P_2^{tissue}}$$

$CTR_{SHI} - CTR_{Har}$ =4.5dB

F. Lin *et al.*, *IEEE IUS*, 2011

Generalization of multi-pulse techniques

Multi-pulse techniques in ultrasound contrast imaging:

- Amplitude modulation (AM), Brock-Fisher *et al,* 1996
- Pulse inversion (PI), Simpson *et al.,* 1999
- Phase coded sequences (PCS), Wilkening *et al.,* 2001
- Contrast pulse sequence (CPS), Phillips, 2001
- Pulse inversion amplitude modulation (PIAM), Eckersley *et al.,* 2005
- Second harmonic inversion (SHI), Pasovic *et al.,* 2011

⋮

⇓

Generalization of multi-pulse techniques: $$r_{sum}(t) = \sum_{k=1}^{K} c_k \left(\sum_{n=1}^{N} a_n b_k^n {q_0}^n(t) \right)$$

Generalization of multi-pulse techniques

Creatis

Single pulse: $p_0(t) = P_0 e^{j\omega_0 t + \varphi_0}$ —In reception→ $r(t) = a_1 q_0(t) + a_2 {q_0}^2(t) + \cdots + a_N {q_0}^N(t) = \sum_{n=1}^{N} a_n {q_0}^n(t)$

Multiple pulses (Suppose k pulses): $p_k(t) = b_k P_0 e^{j\omega_0 t + \varphi_0}$ —Weighted sum in reception→ $r_{sum}(t) = \sum_{k=1}^{K} c_k \left(\sum_{n=1}^{N} a_n b_k^n {q_0}^n(t) \right)$

$$r_{sum}(t) = \sum_{n=1}^{N} \left(\sum_{k=1}^{K} c_k b_k^n \right) a_n {q_0}^n(t)$$

$\left(\sum_{k=1}^{K} c_k b_k^n \right)$: the weight of the n[th] component in the final summed signal

Example:

PI: $\begin{cases} b = (1,-1) \\ c = (1,1) \end{cases}$ → $r_{sum}(t) = \sum_{n=1}^{N} (1 + e^{jn\pi}) a_n {q_0}^n(t)$ —N=3→ $r_{sum}(t) = 0 + 2a_2 {q_0}^2(t) + 0$

AM: $\begin{cases} b = \left(1, \frac{1}{2}\right) \\ c = (1,-2) \end{cases}$ → $r_{sum}(t) = \sum_{n=1}^{N} (1 - 2^{1-n}) a_n {q_0}^n(t)$ —N=3→ $r_{sum}(t) = 0 + 0.5a_2 {q_0}^2(t) + 0.75a_3 {q_0}^3(t)$

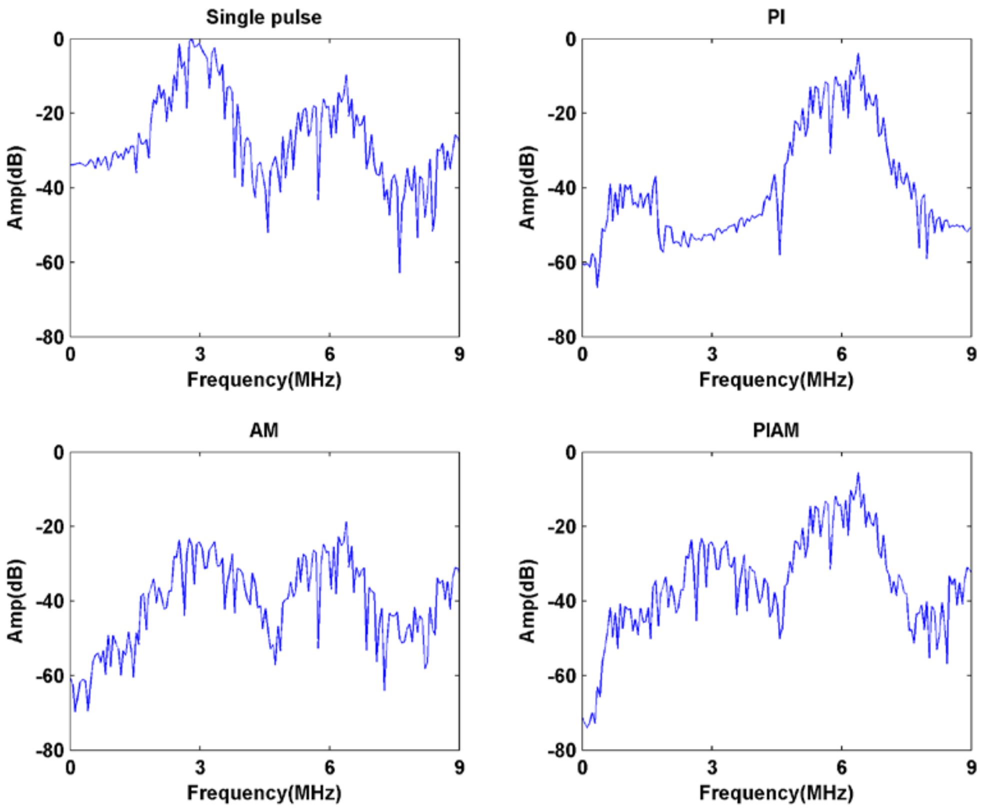
PIAM: $\begin{cases} b = \left(1, \frac{e^{j\pi}}{2}\right) \\ c = (1,2) \end{cases}$ → $r_{sum}(t) = \sum_{n=1}^{N} (1 + 2^{1-n} e^{jn\pi}) a_n {q_0}^n(t)$ —N=3→ $r_{sum}(t) = 0 + 1.5a_2 {q_0}^2(t) + 0.75a_3 {q_0}^3(t)$

F. Lin *et al.*,

Generalization of multi-pulse techniques

Simulation: spectra from the single pulse, PI, AM, PIAM sequences in the UCA region

$f_0 = 3$ MHz



PI:
$$r_{sum}(t) = 0 + 2a_2 {q_0}^2(t) + 0$$

AM:
$$r_{sum}(t) = 0 + 0.5a_2 {q_0}^2(t) + 0.75a_3 {q_0}^3(t)$$

PIAM:
$$r_{sum}(t) = 0 + 1.5a_2 {q_0}^2(t) + 0.75a_3 {q_0}^3(t)$$

Recall: *odd-order components create spectrum components at odd multiples of ω_0, while even-order nonlinear components create spectral components at even multiples of ω_0.*

- The simulation results agree with the experimental results presented in previous literature [Eckersley, 2005].
- The generalized equation is validated by the simulation.

F. Lin *et al.*,

Outline

- **Basic of** (Linear) **ultrasound and medical ultrasound imaging**
- **Advanced in** (Nonlinear) **Ultrasound**
  - Ultrasound Contrast Agent
  - Nonlinear ultrasound
- **Improvement of ultrasound imaging or Nonlinear imaging**
  - Harmonic imaging
  - Multipulses imaging or Contrast (nonlinear) imaging
    - Pulse inversion (PI)
    - Second Harmonic Inversion (SHI)
  - Generalization of multi-pulse techniques
  - Influence of scatterer motion to phased multipulses method
- **CREANUIS: Simulation of nonlinear ultrasound images**

Creatis

Influence of bubbles motion

However, during the experimental evaluation to SHI technique, the reduction of second-harmonic amplitude of bubble responses was sometimes observed.

Is the preservation of bubbles second harmonic related to bubbles motions?

$$RF_{sum} = RF_1(\varphi = 0) + RF_2(\varphi = \frac{\pi}{2} + \frac{4\pi\Delta z}{\lambda})$$

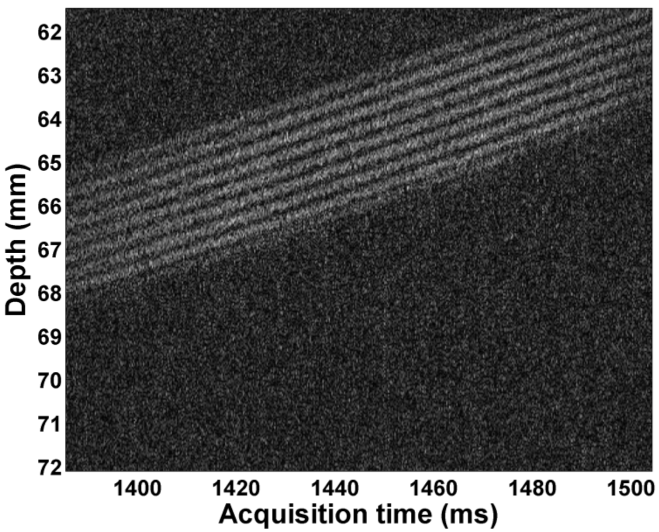
Δz: axial bubbles motion
λ: transmitted wavelength

Influence of scatterer motion to phased multipulses method

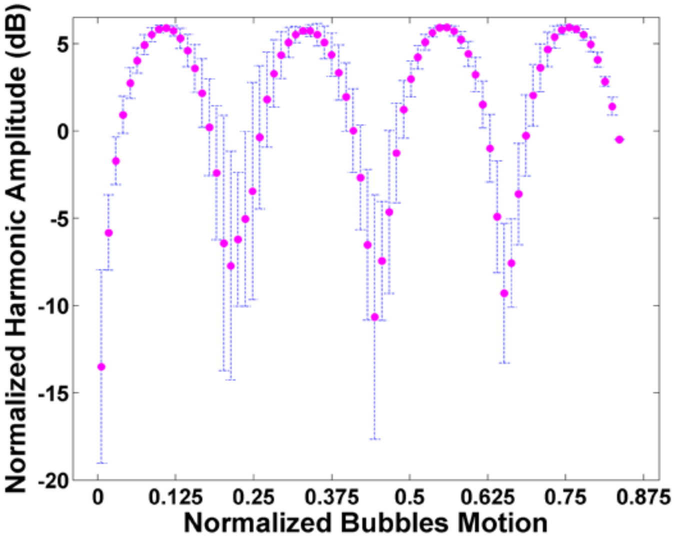
Influence of bubbles motion

***In-vitro* experimental results**

M mode image



Second harmonic amplitude of SHI versus bubbles motion



Bubbles motion is normalized to transmitted wavelength

$$RF_{sum} = RF_1(\varphi = 0) + RF_2(\varphi = \frac{\pi}{2} + \frac{4\pi\Delta z}{\lambda})$$

Δz: axial bubbles motion

λ: transmitted wavelength

Maximum at: $\Delta z = (2i+1)\lambda/8$ and $\varphi = \pi \Rightarrow PI$

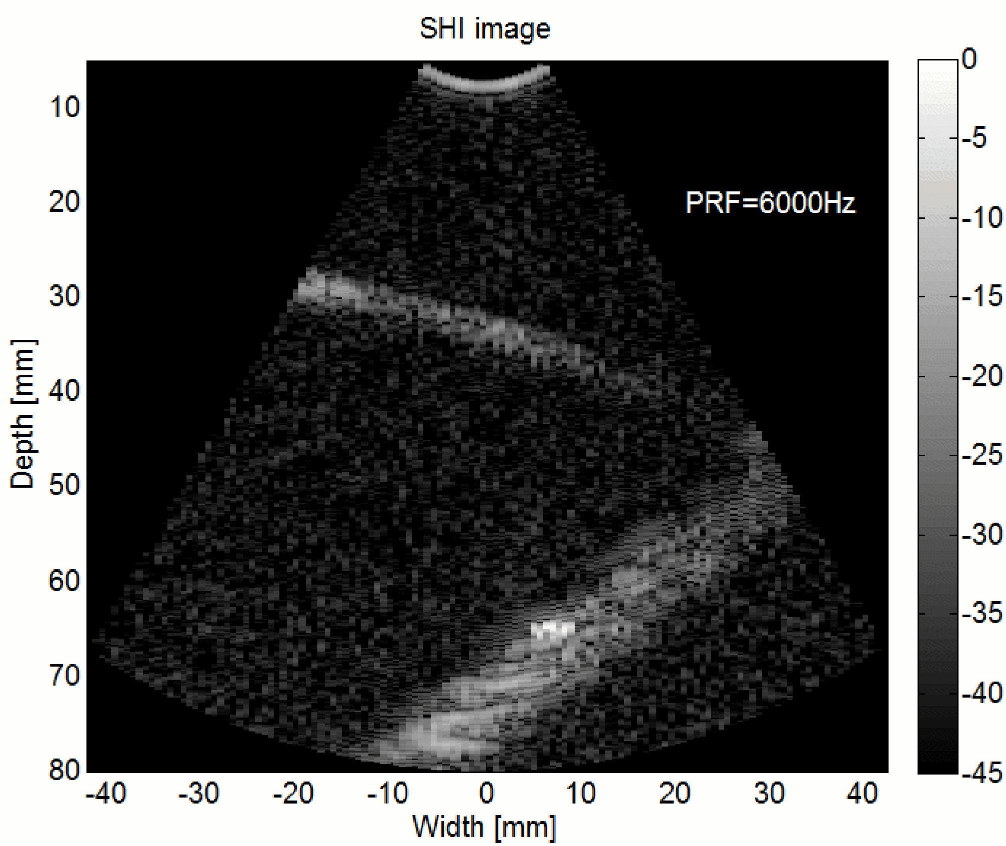
Minimum at: $\Delta z = 2i\,\lambda/4$ and $\varphi = \pi/2 \Rightarrow SHI$

F. Lin *et al.*, *IEEE UFFC*, 2013

Influence of scatterer motion to phased multipulses method

Optimization of SHI

Update PRF according to CTR until the optimal PRF is found



Outline

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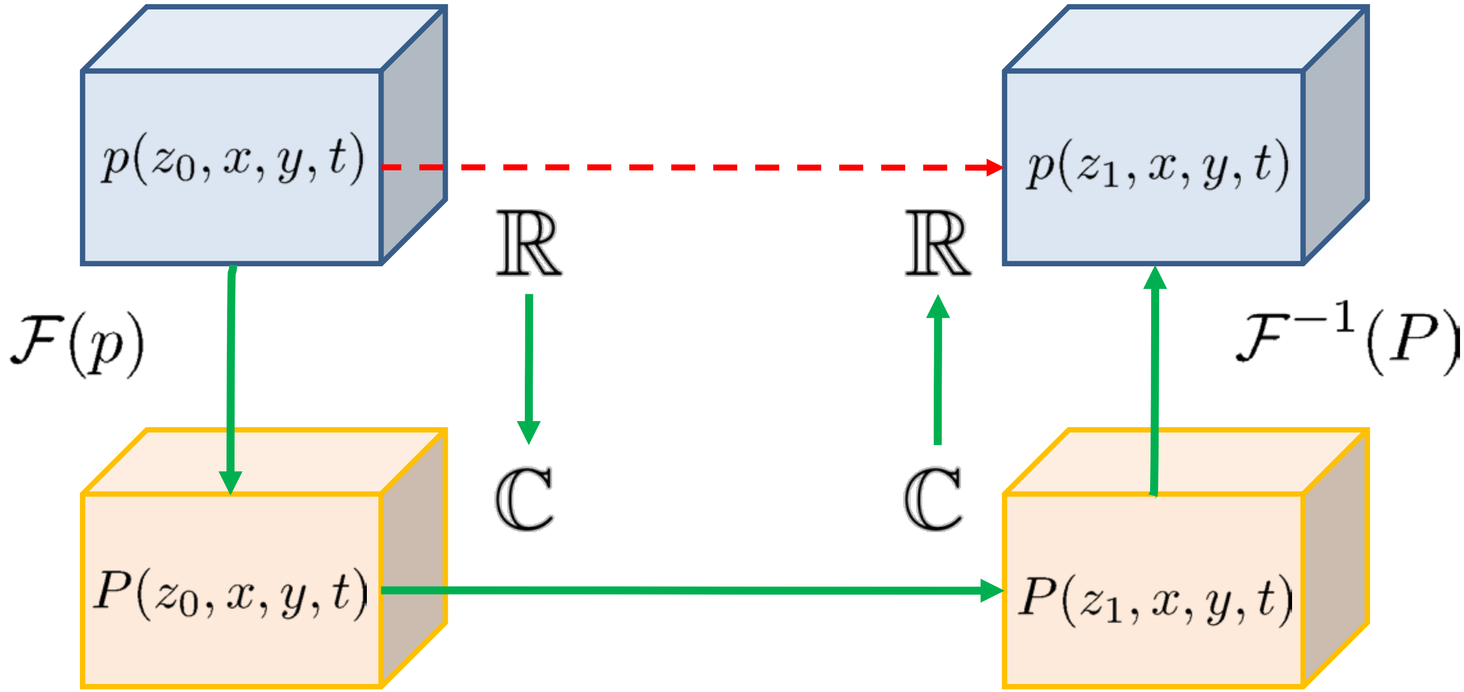
- **CREANUIS: Simulation of nonlinear ultrasound images**

contrast agent imaging, new non linear imaging, motion estimation, segmentation, …

- **Nonlinear propagation**
  - Various solutions : finite difference methods, angular spectrum methods (ASM), coupled approaches
  - Long computation time (easily up to hours)
  - Homogeneous nonlinear coefficient
- **Ultrasound image simulation**
  - Mainly based on linear propagation
  - Sum and delay algorithm

ASM principle



$$\mathcal{F}(p) : p(z, x, y, t) \rightarrow P(z, f_x, f_y, f_t) = \int\int\int p(z, x, y, t) e^{-i2\pi(f_x x + f_y y - f_t t)} dxdydt$$

Property:
$$\begin{cases} \mathcal{F}\left(\frac{\partial^n p}{\partial v^n}\right) = (-2i\pi f_v)^n \mathcal{F}(p), \quad v = x \ or \ y \\ \mathcal{F}\left(\frac{\partial^n p}{\partial t^n}\right) = (2i\pi f_t)^n \mathcal{F}(p) \end{cases}$$

Generalized ASM (GASM)

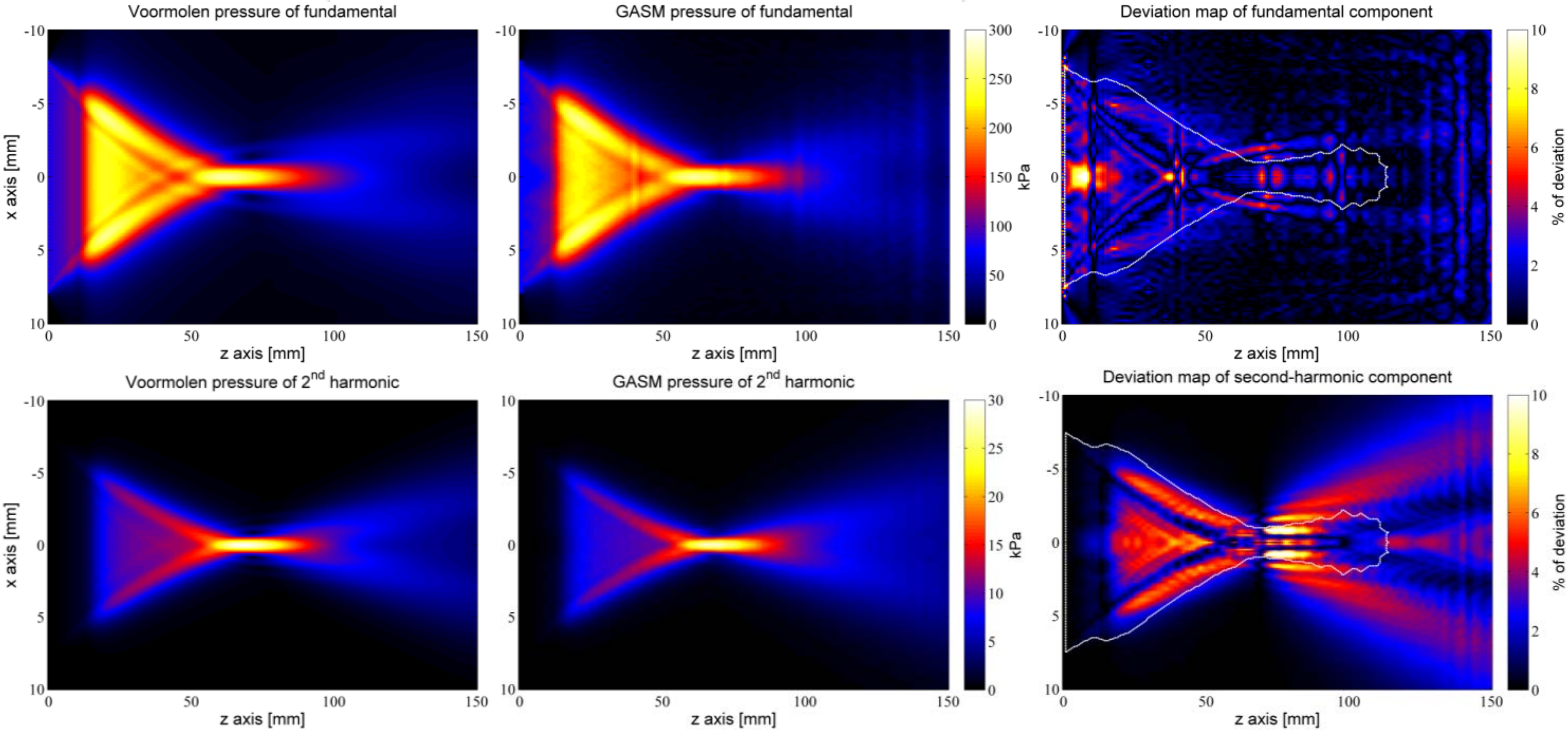
$$\begin{cases} p_1(z,x,y,t) = \mathcal{F}^{-1}\left(P_0(z_0,f_x,f_y,f_t)e^{iK(z-z_0)}\right) \\ p_2(z,x,y,t) = \mathcal{F}^{-1}\left(\frac{-ik_t^2}{2K\rho_0 c_0^2}\int_{z_0}^{z}\mathcal{F}(\beta p_1^2)e^{-iKu}du \times e^{iKz}\right) \end{cases}$$

With $\begin{cases} P_0 & \text{is the FT of the transmitted US wave} \\ K & \text{is the 3D } k\text{-vector} \end{cases}$

In power attenuating media (α_0, γ) [Szabo (1978)]

$$K_a(k_x,k_y,k_t) = K(k_x,k_y,k_t) - i\alpha_0\left(\frac{f_t}{1e6}\right)^{\gamma}$$

Simulation results

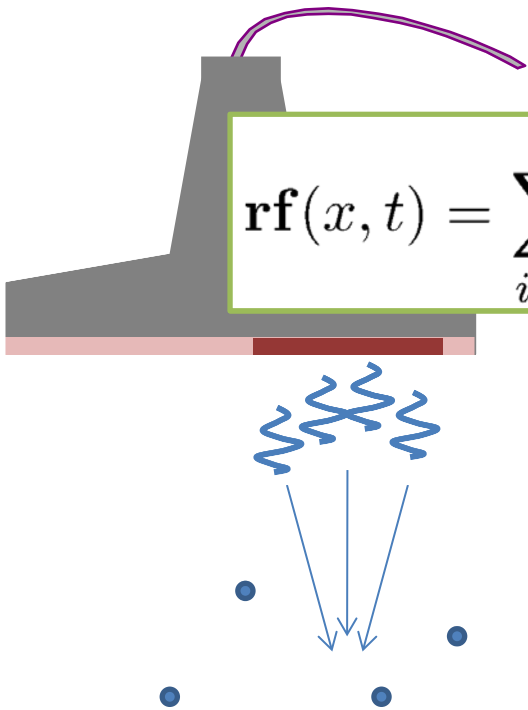


$$D_i = \frac{|p_i^{Ref} - p_i^{GASM}|}{\max(p_i^{Ref})}$$

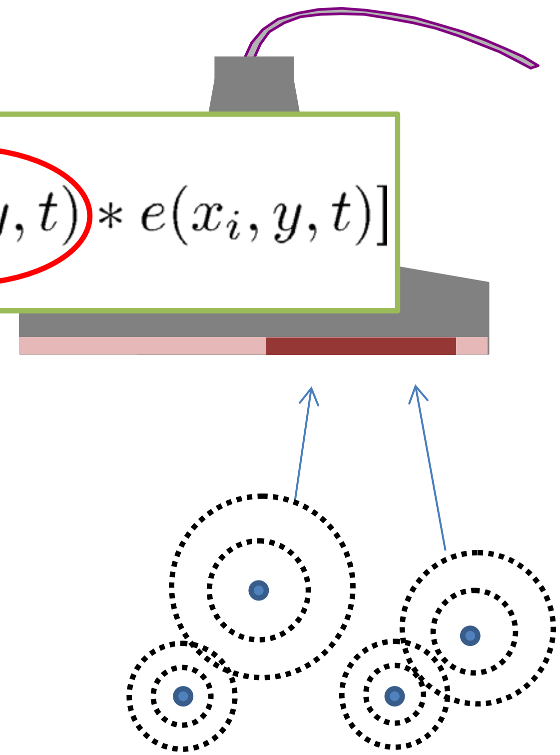
	Mean deviation	Maximal deviation
D_1	3.7 %	8.1 %
D_2	3.4 %	7.8 %

Transmission – **GASM**

Reception – **CREANUIS**
(**CREA**tis **N**onlinear **U**ltrasound **I**mage **S**imulation)



$$\mathbf{rf}(x,t) = \sum_{i=1}^{N} w_r(x_i)\,[h(x_i,y,t) * e(x_i,y,t)]$$



The pressure field is known thanks to the GASM simulations

+

The reception is handled as in FieldII: spatial impulse response, apodization, focalization…

Medium

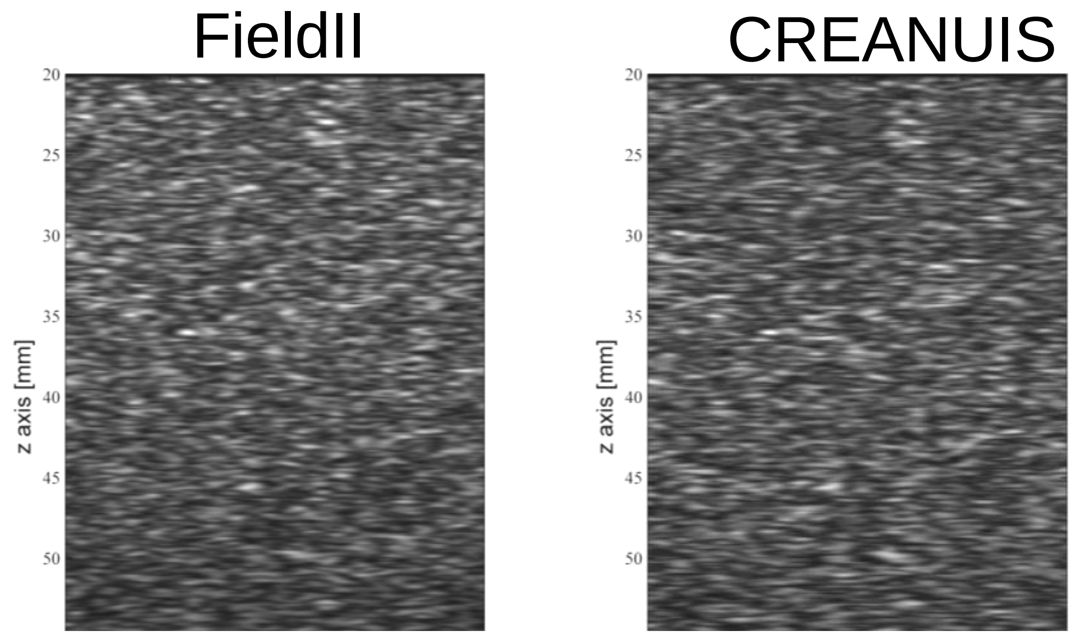
- 30 scatterers/mm^3

TX signal

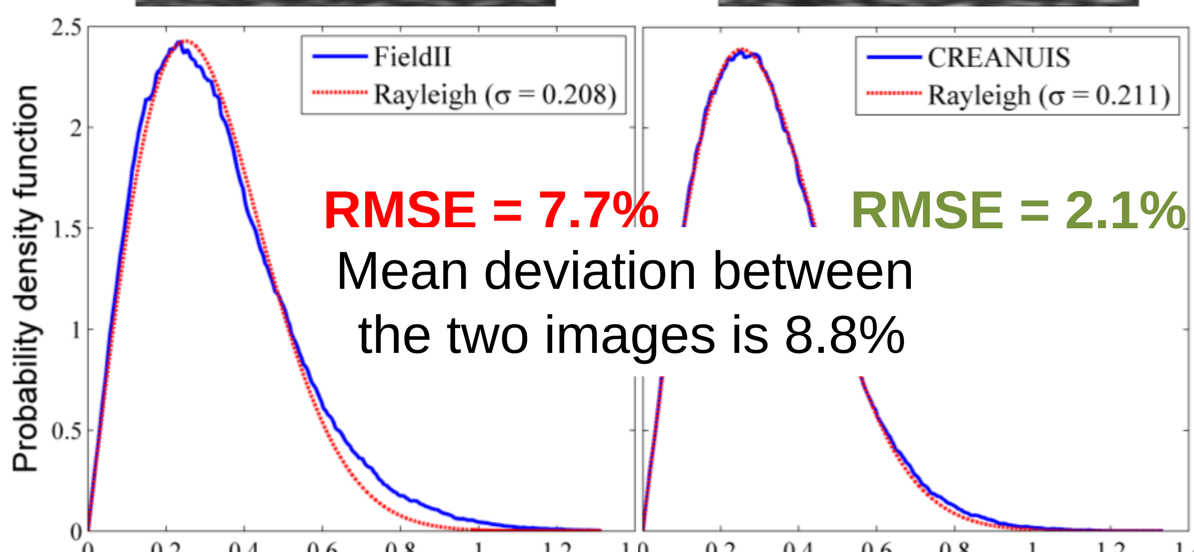
- 3-cycle sine at 5 MHz
- Hanning windows
- Focalization at 40 mm

Beamforming

- Hanning apodization in TX and RX



Theoretical probability density function: Rayleigh distribution



RMSE = 7.7%

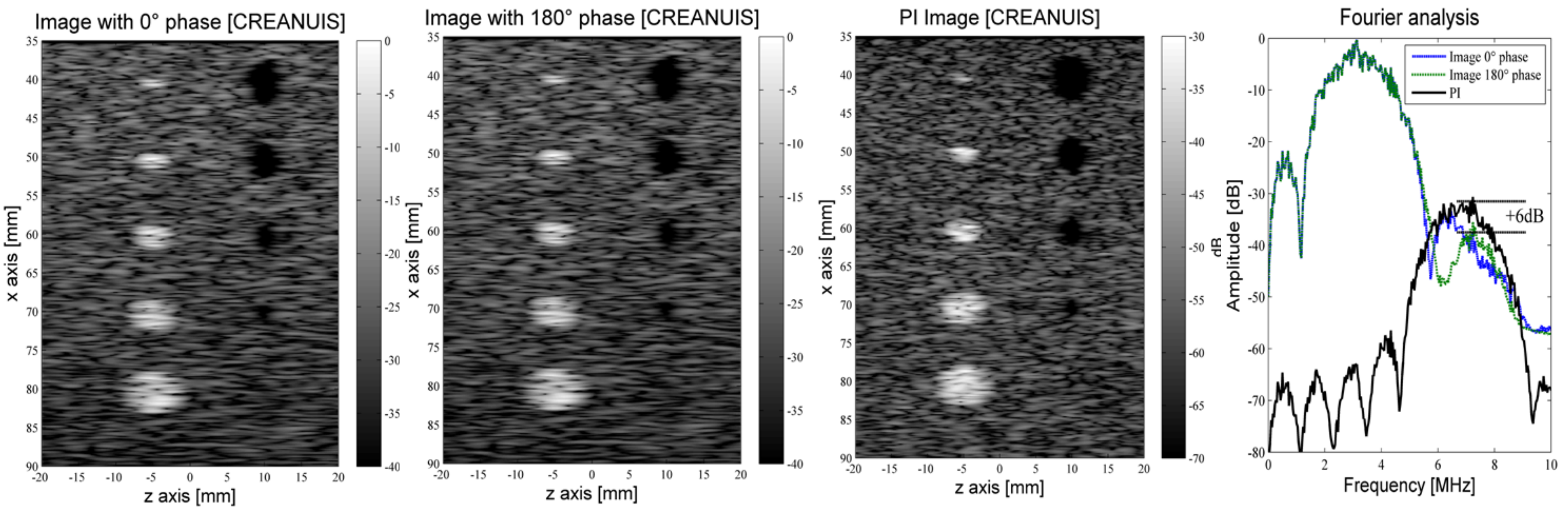
RMSE = 2.1%

Mean deviation between the two images is 8.8%

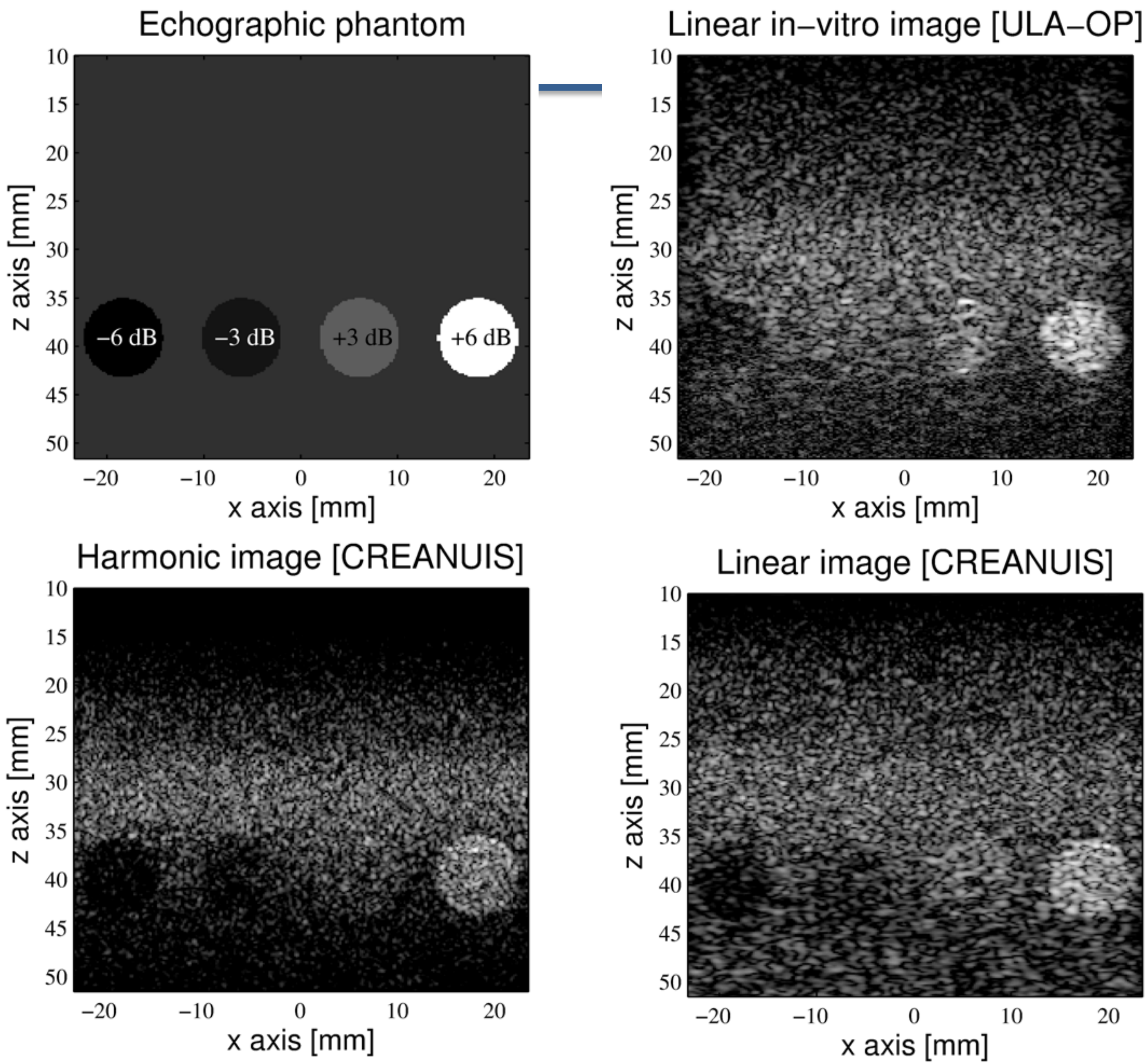
CREANUIS – New simulation modalities

Nonlinear imaging:

- Amplitude modulation
- Second harmonic inversion
- **Pulse inversion**
- …



Varray *et al.*, UMB 2013

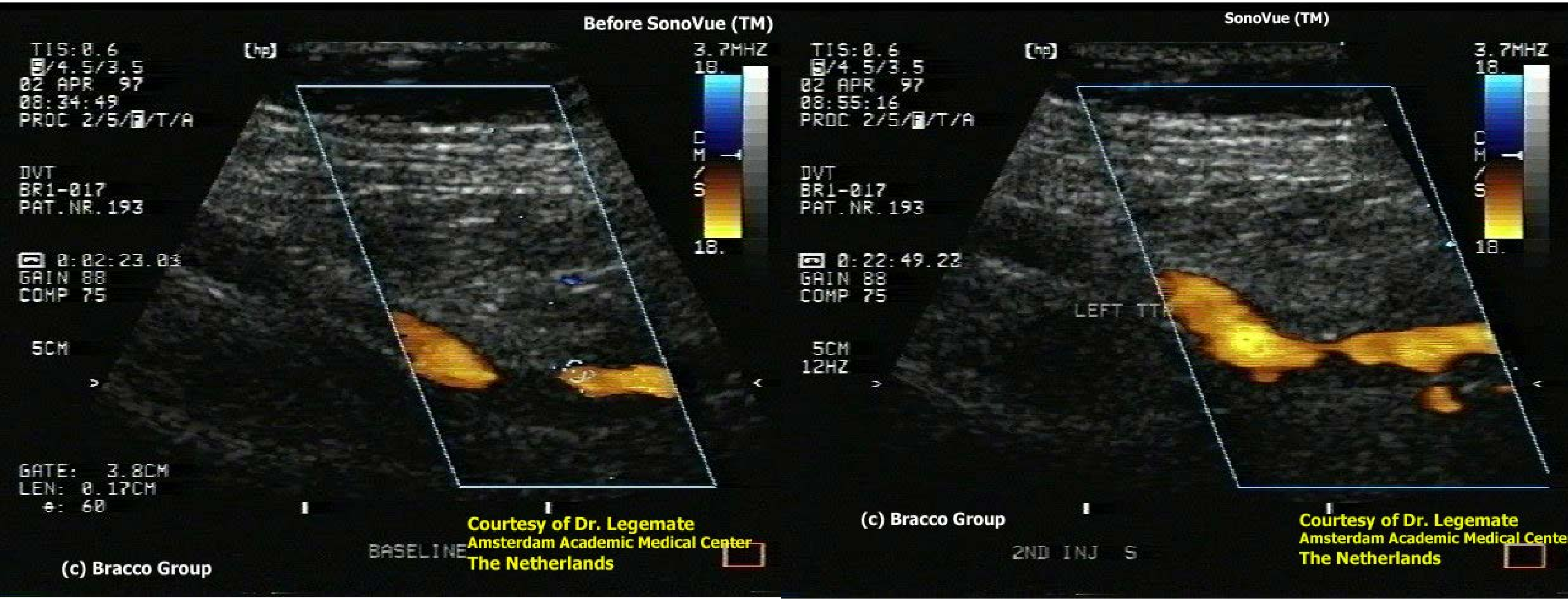


CREANUIS is avalaible : https://www.creatis.insa-lyon.fr/site/fr/CREANUIS

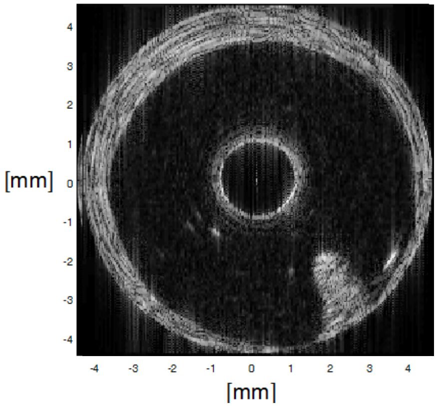
Peripheral Vessels disease

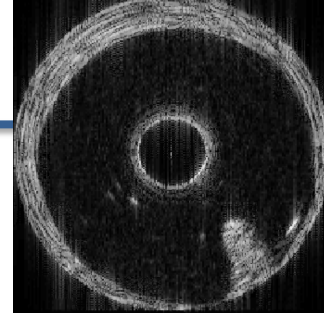
Before SonoVue

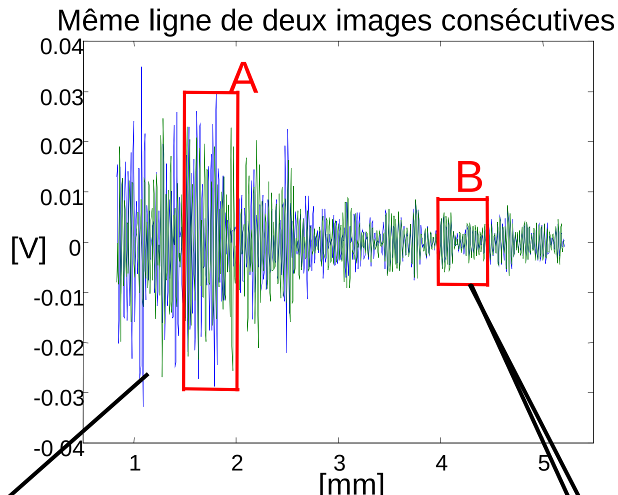
After SonoVue

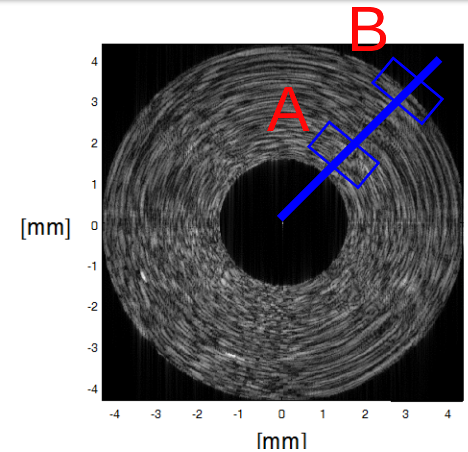


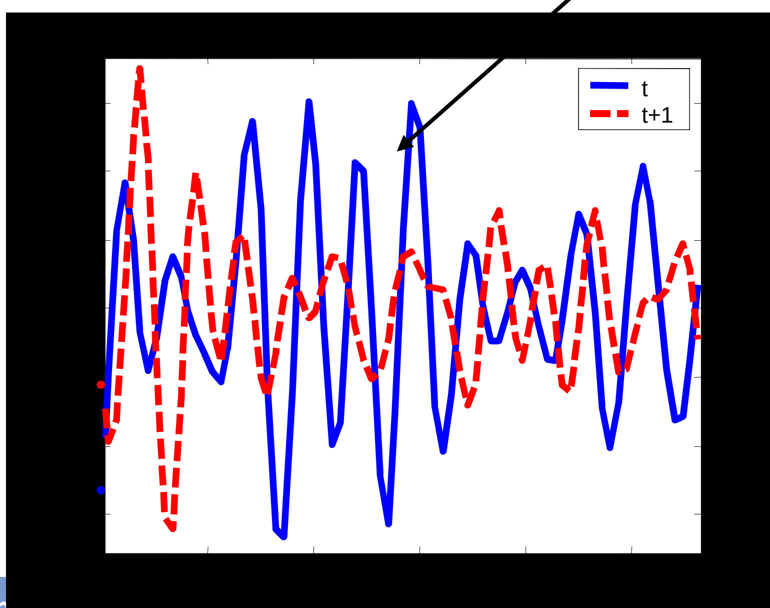
Uncorrelation

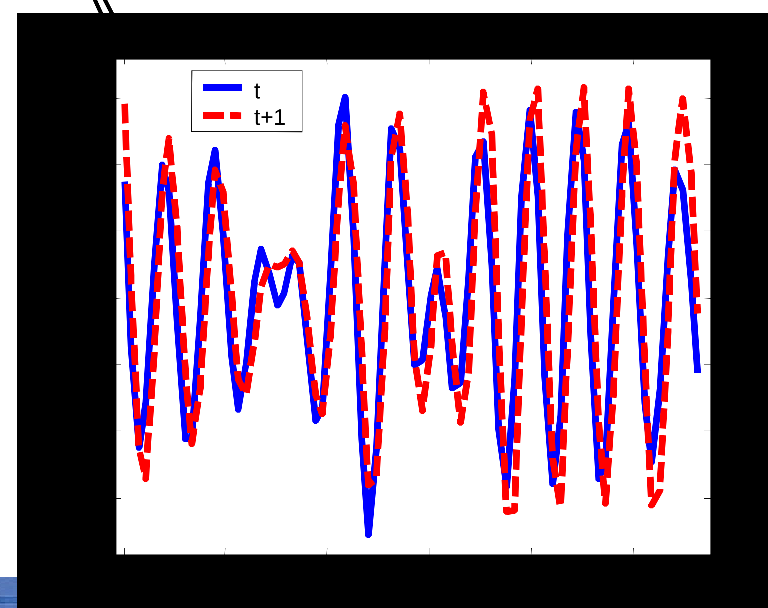




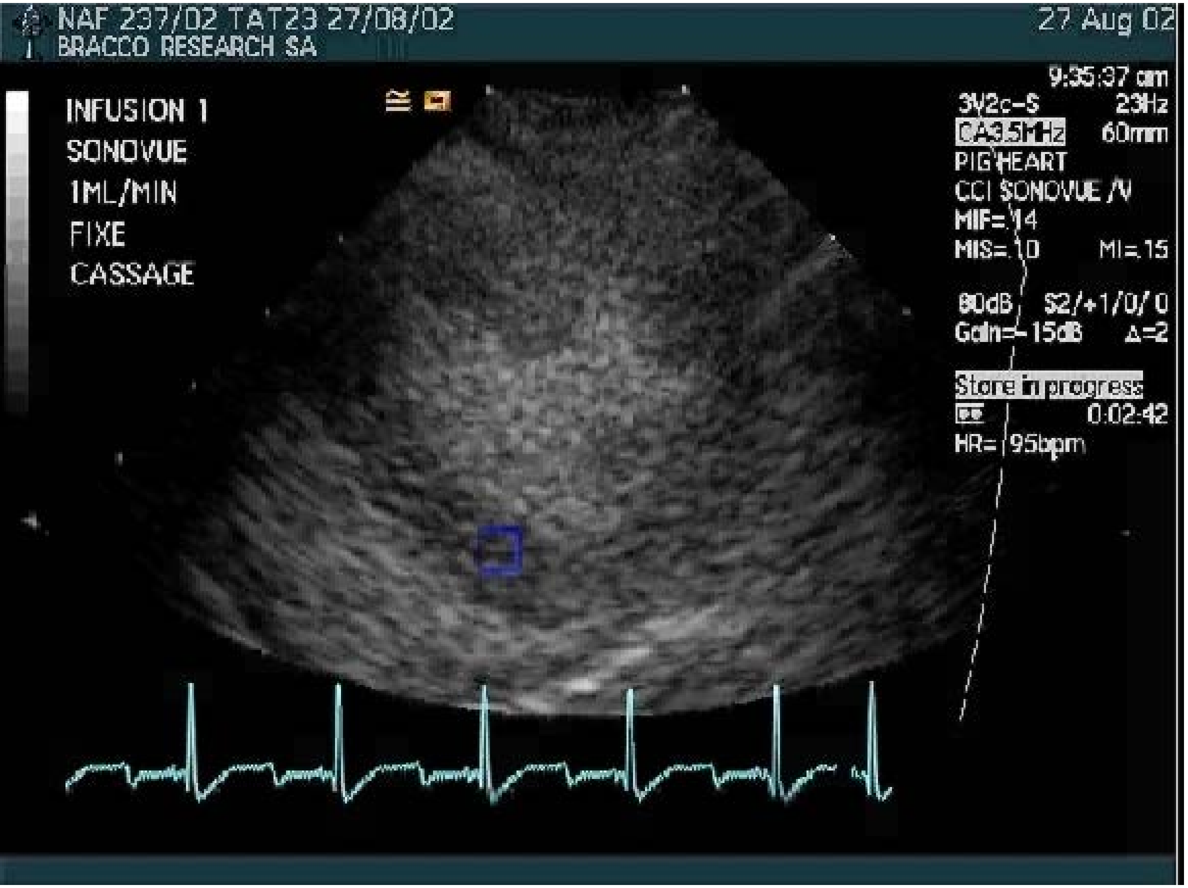


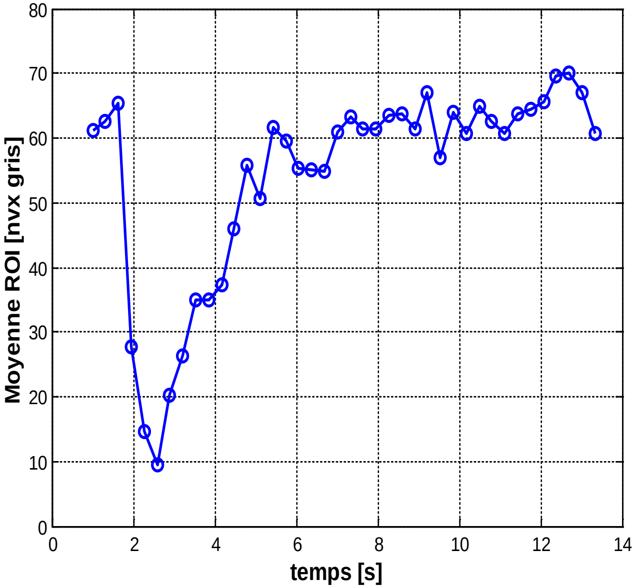






Destruction and reperfusion mode
Quantification of perfusion





One shot at high intensity to destroy the contrast agent

Imaging dedicated to contrast agent

higher $\dfrac{\text{signal}}{\text{noise}}$

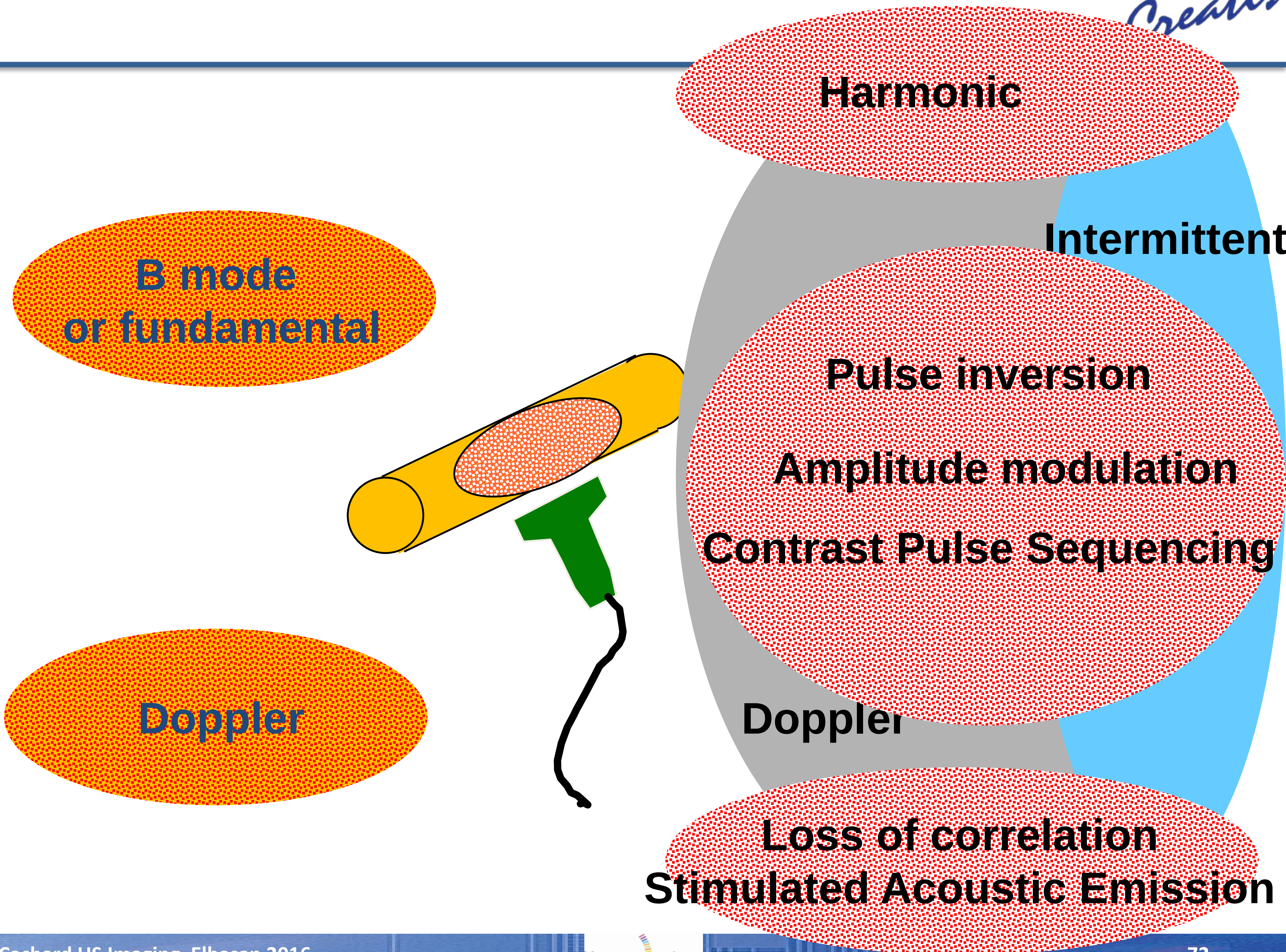
signal ← wave scattered by contrast agent

noise ← wave scattered by tissue

One scan

Two or more scans

Transmitted pulses and signal processing


Harmonic
Intermittent
B mode
or fundamental
Pulse inversion
Amplitude modulation
Contrast Pulse Sequencing
Doppler
Doppler
Loss of correlation
Stimulated Acoustic Emission