Alternative linear method for simulation of US data sets:

COLE

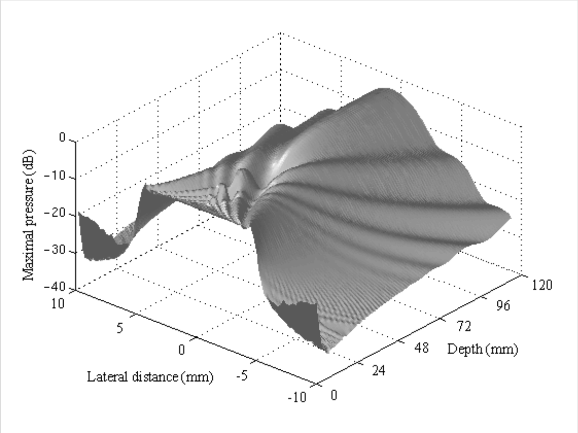
Hang Gao, Jan D’hooge

Cardiovascular Imaging and Dynamics
Medical Imaging Research Center
Catholic University of Leuven
Leuven, Belgium

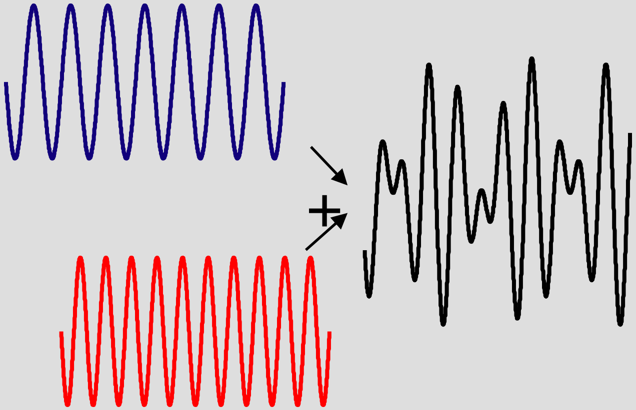
Trondheim, May 4th, 2010

Ultrasound wave propagation

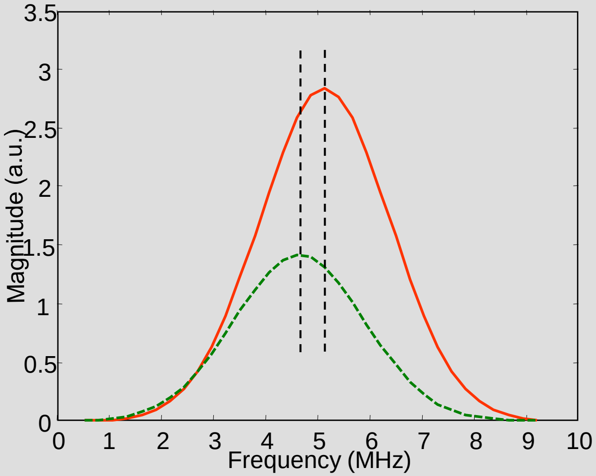
- Diffraction



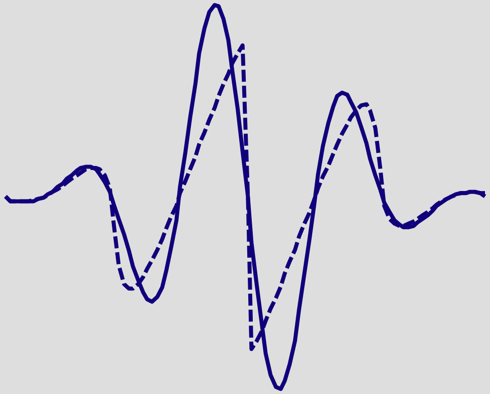
- Dispersion



- Attenuation

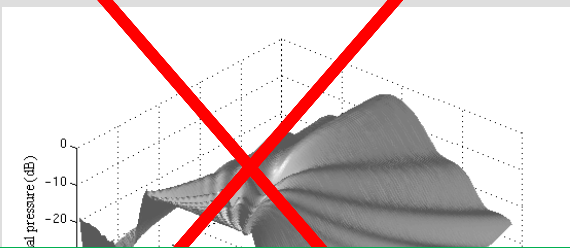


- Non-linearity

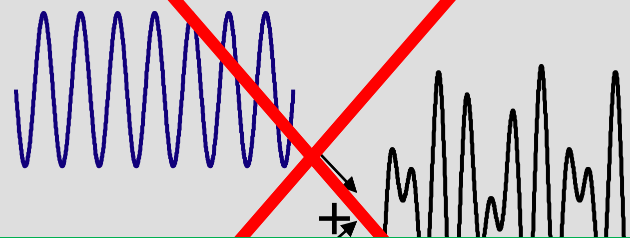


Ultrasound wave propagation

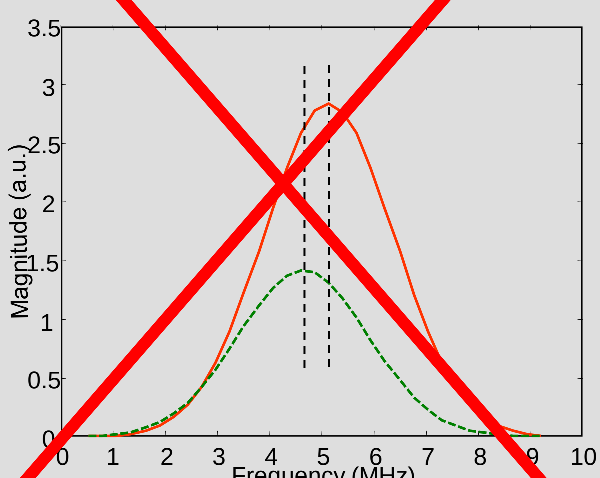
- Diffraction

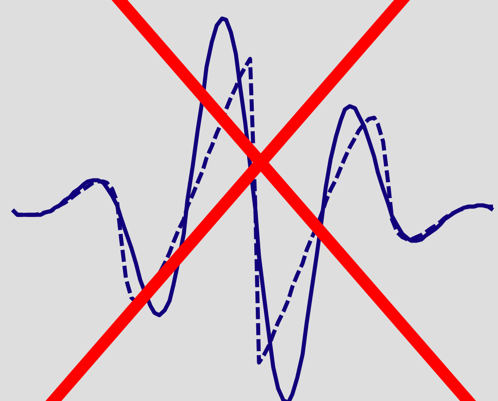


- Dispersion



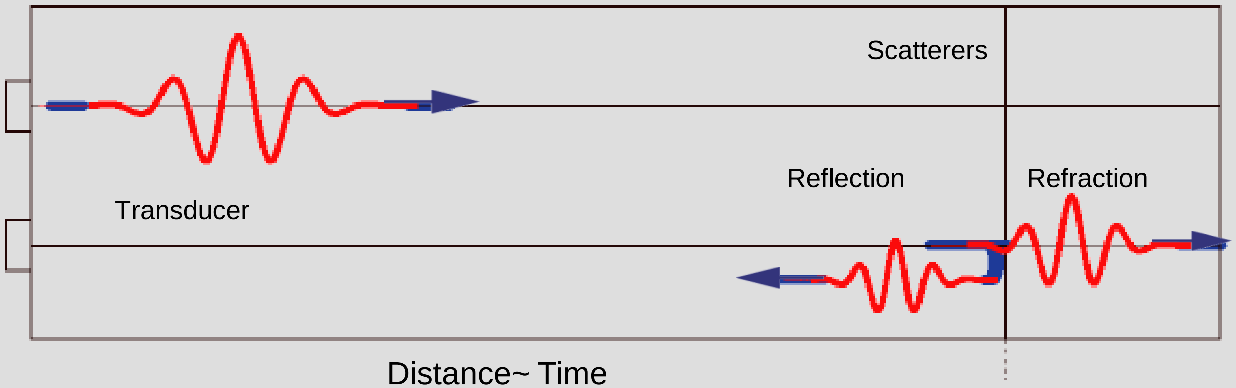
Point Spread Function (PSF) is spatially invariant





Convolution model

Received signal = delayed and scaled replica of the transmitted pulse



$$S(t) = a.P(t - 2d/c) = P(t) \otimes a.\delta(t - 2d/c)$$

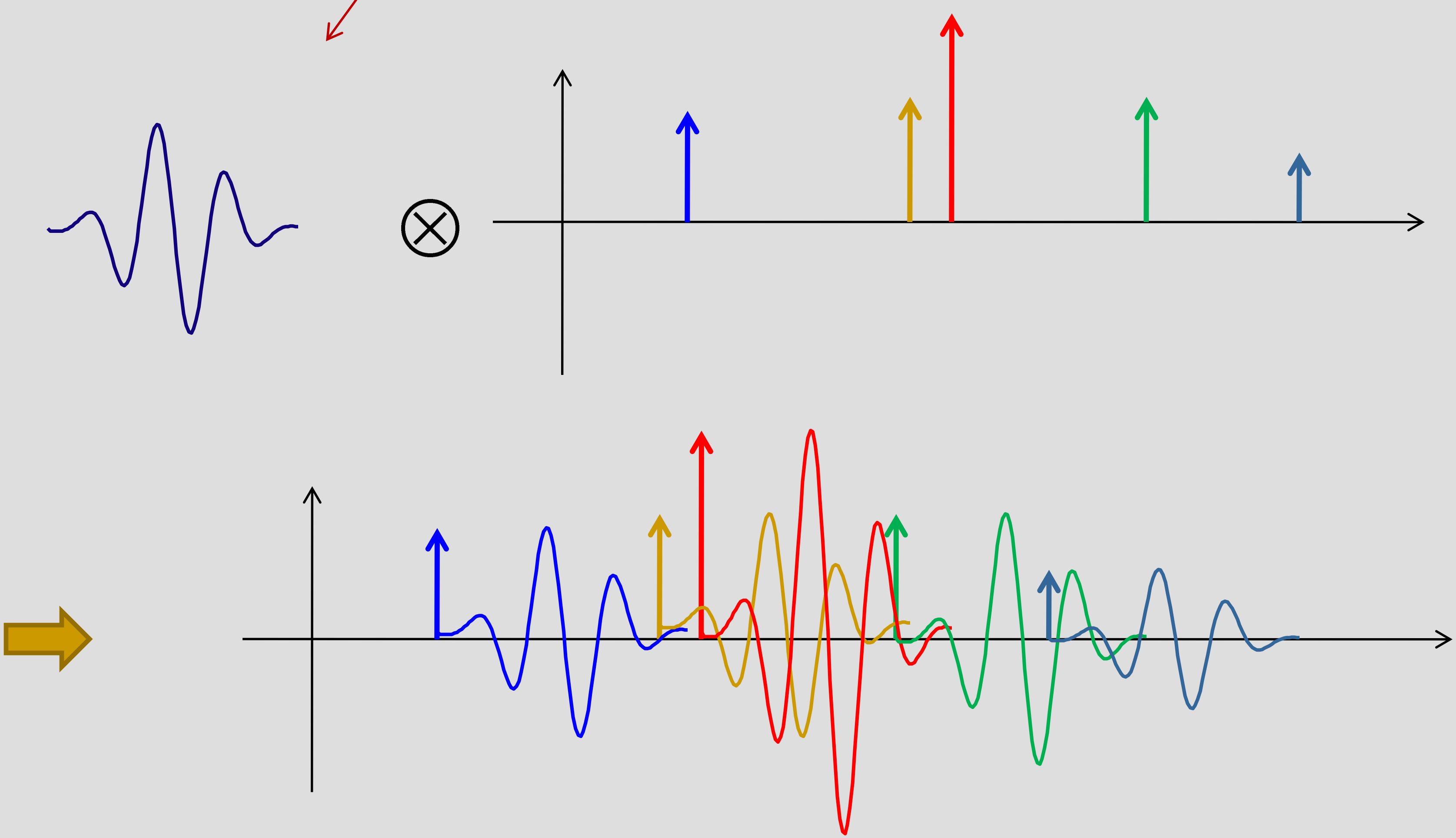
$$S(t) = P(t) \otimes \sum_i a_i\, \delta(t - 2d_i/c)$$

OR $$S(x) = P(x) \otimes \sum_i a_i\, \delta(x - x_i)$$ *(spatial domain)*

Convolution model graphically

$$S(x) = P(x) \otimes \sum_i a_i\, \delta(x - x_i)$$

Tissue scatterer function T(x)



Convolution model in 2D

$$I(x,y) = \iint T(x',y') \times H(x-x', y-y')dx'dy'$$

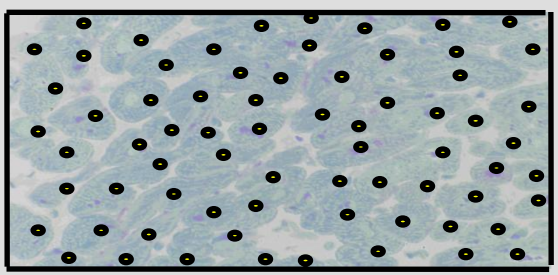
with

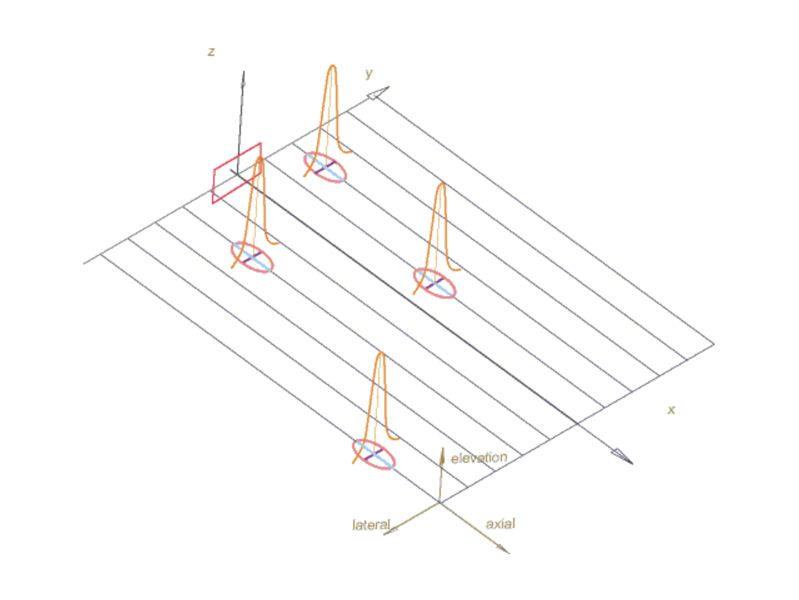
$$T(x,y) = \sum_q a_q \delta(x-x_q, y-y_q)$$

and

$$H(x,y) = \exp[-(x^2/\sigma_x^2 + y^2/\sigma_y^2)/2]\cos(2\pi fx)$$

Discrete scatterer model: T(x,y)



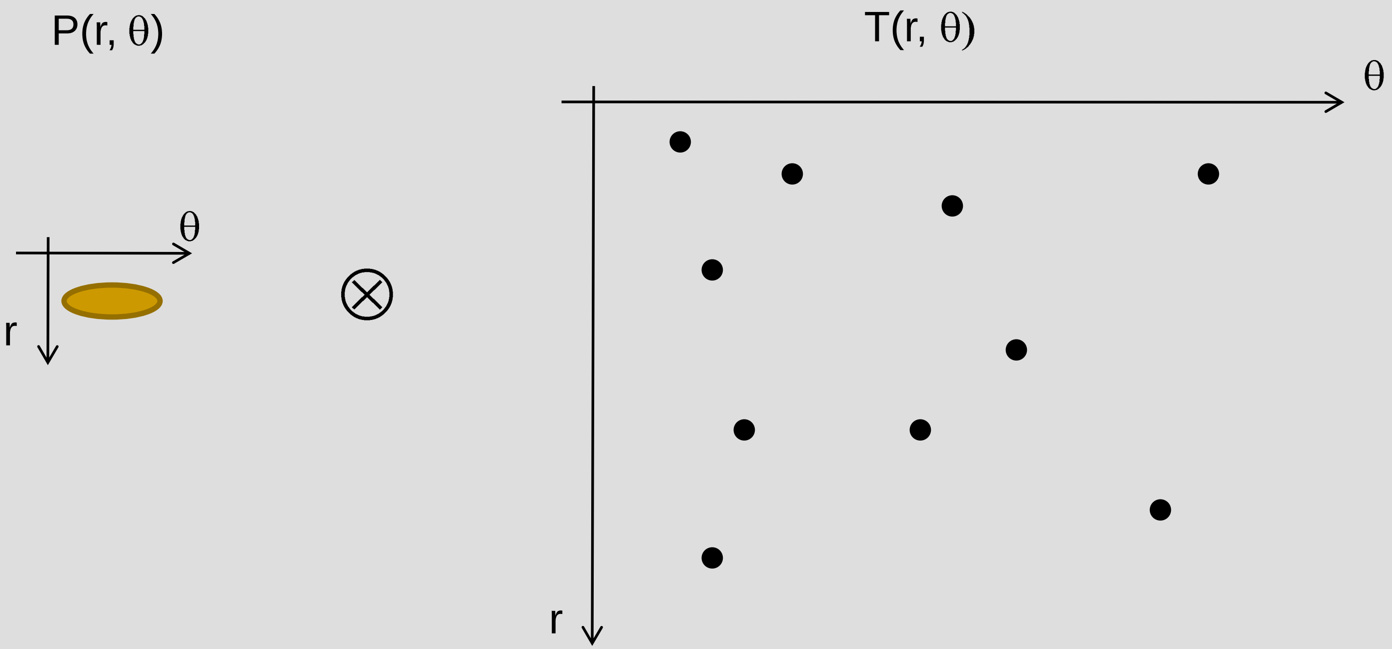


2D sector images?

$$S(r,\theta) = P(r, \theta) \otimes T(r, \theta) = P(r, \theta) \otimes \sum_i a_i \, \delta(r - r_i, \theta - \theta_i)$$

Problem:

- Once $P(r,\theta)$ is defined the effective PSF becomes spatially variant through the convolution process as image lines diverge

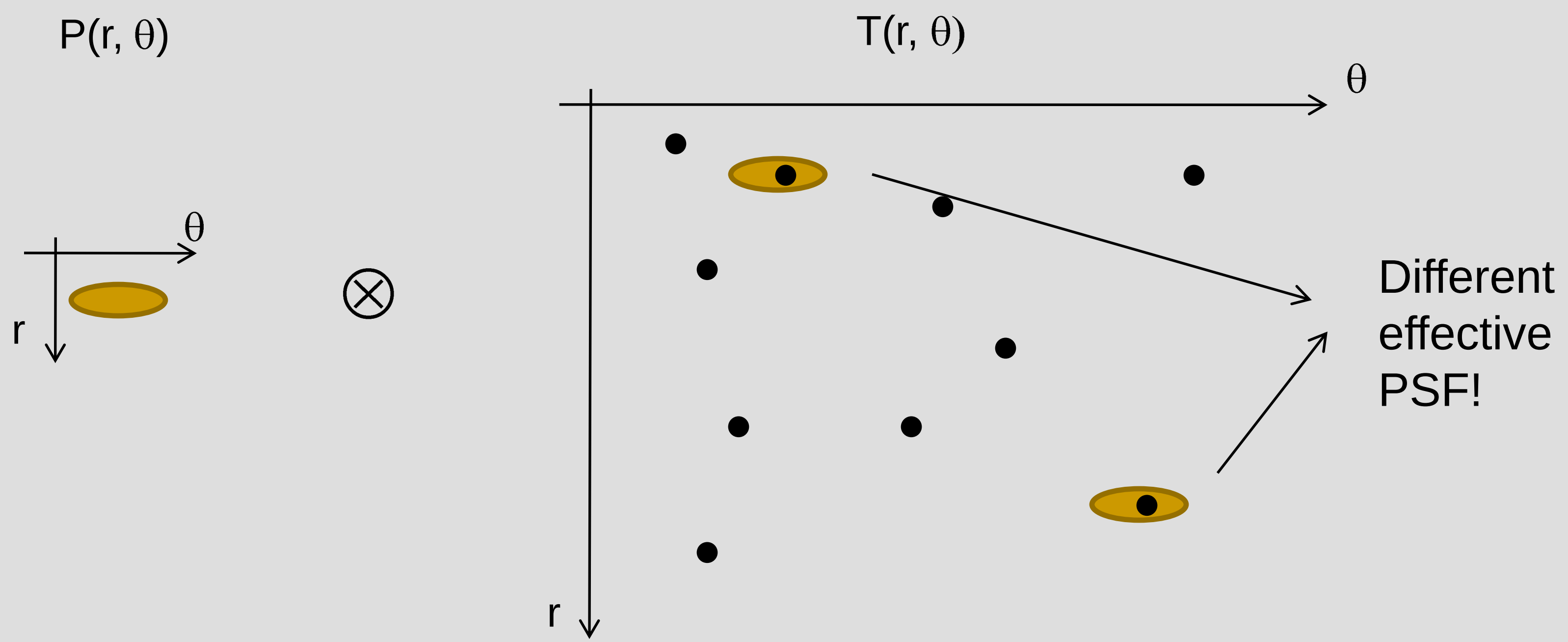


2D sector images?

$$S(r,\theta) = P(r,\theta) \otimes T(r,\theta) = P(r,\theta) \otimes \sum_i a_i \, \delta(r - r_i, \theta - \theta_i)$$

Problem:

- Once $P(r,\theta)$ is defined the effective PSF becomes spatially variant through the convolution process as image lines diverge



Convolution model in 3D

⇨ Straight forward extension

However:

• 3D convolution can require a high memory load / high computational load especially when the convolution grid is fairly dense (to have good accuracy)

COLE

Do not take the perspective of the scatterers but rather of the ultrasound wave

Part of the wave will be reflected if the scatterer is within the beam

→ central in the beam: contribute signicantly

→ far from the center of the beam: contribute little

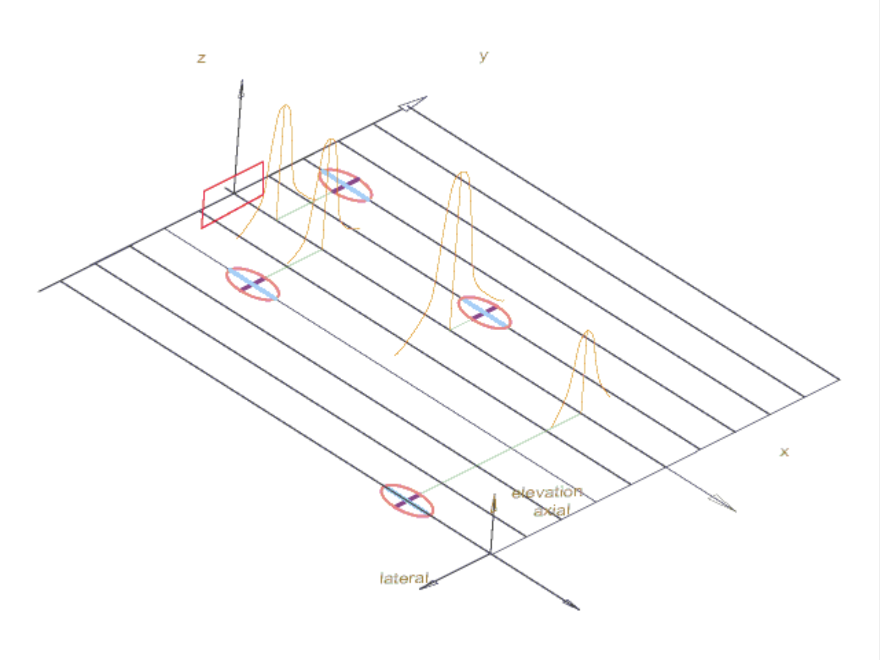
Weigh the contribution of the scatterer with the distance from the image line

$$I(r, \theta_m, \varphi_n) = H(r) \otimes T'(r, \theta_m, \varphi_n)$$

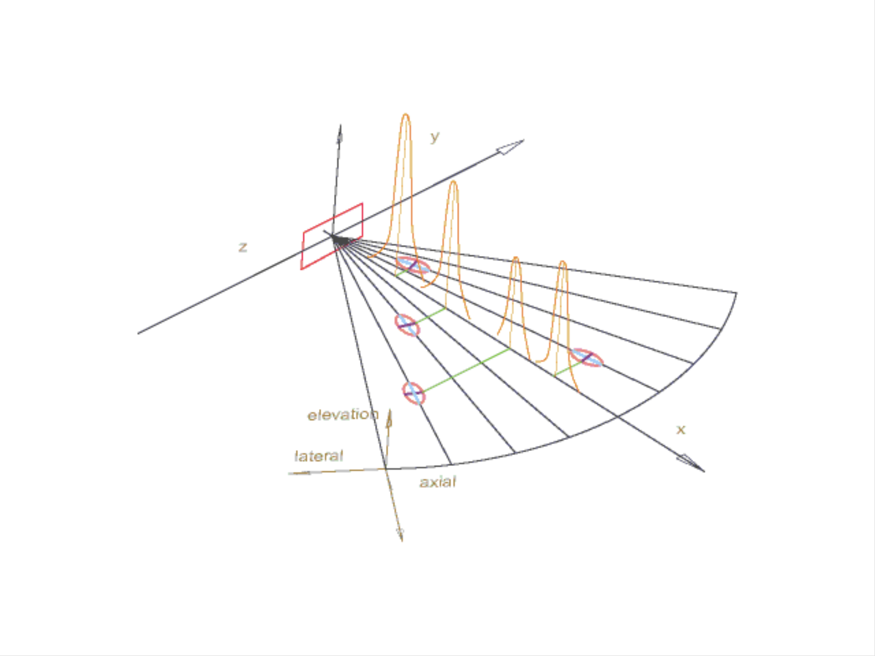
$$T'(r, \theta_m, \varphi_n) = \sum_{q=1}^{N} w_q a_q \delta(r - r_q),$$

COLE graphically

Cartesian



Polar



COLE: practical

$$I(r,\theta_m,\varphi_n) = H(r) \otimes T'(r,\theta_m,\varphi_n)$$

$$T'(r,\theta_m,\varphi_n) = \sum_{q=1}^{N} w_q a_q \delta(r - r_q),$$

How to define the weighting w_q ?

- **from an analytical expression of the beam profile:**

  e.g.the Gaussian PSF defined as

  $$w_q = \exp\left\{-[r_q^2(\theta_q - \theta_m)^2 / \sigma_L^2 + r_q^2(\varphi_q - \varphi_n)^2 / \sigma_E^2)]/2\right\}$$

- **from a simulated beam profile look up table (LUT)**

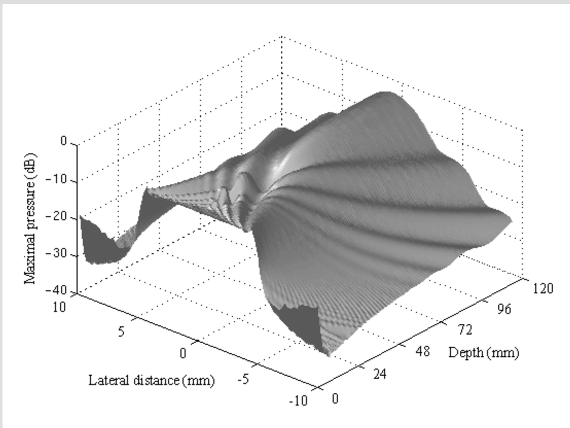
  by its axial distance r_q, its lateral angle $\theta_q - \theta_m$ and its elevation angle $\varphi_q - \varphi_n$ (from scatterer 'q' to the image line);

- **from a measured beam profile LUT by the same indices**

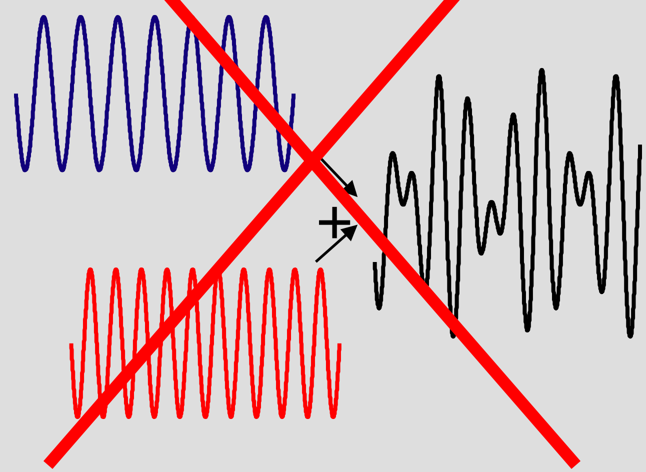
  (i.e r_q $\theta_q - \theta_m$and $\varphi_q - \varphi_n$) .

COLE with LUT

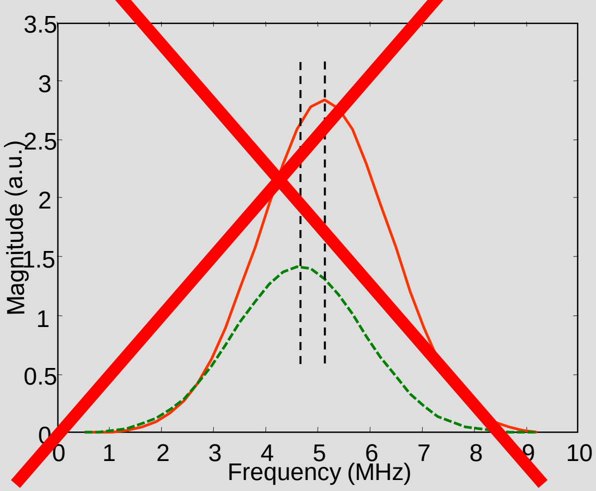
- Diffraction



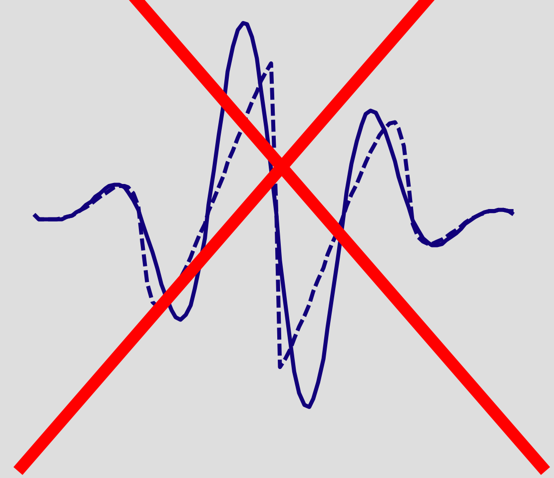
- Dispersion



- Attenuation



- Non-linearity



Methods

Phantom:

20 Scatterers were positioned at equidistant positions in front of the transducer within a background of scattering noise.

Simulation Parameters

- **Gaussian PSF (σL = 1.33 mm)**
- **Beam Profile LUT**

The LUT (a) is tabulated from the realistic beam profile (b) produced by a 64 crystal phased array transducer measuring 10x14 mm and transmitting a 2.5 MHz Gaussian pulse with a -6 dB relative bandwidth of 60%, with fixed transmit and receive foci at a depth of 60 mm using “Field II”

Transducer frequency	2.5 MHz
Sampling frequency	50 MHz
Ultrasound velocity	1500 m/s
Tilt Angle	0 degree
Start Depth	10 mm
Depth	100 mm
Size of ROI	60 degree
Number of lines	60

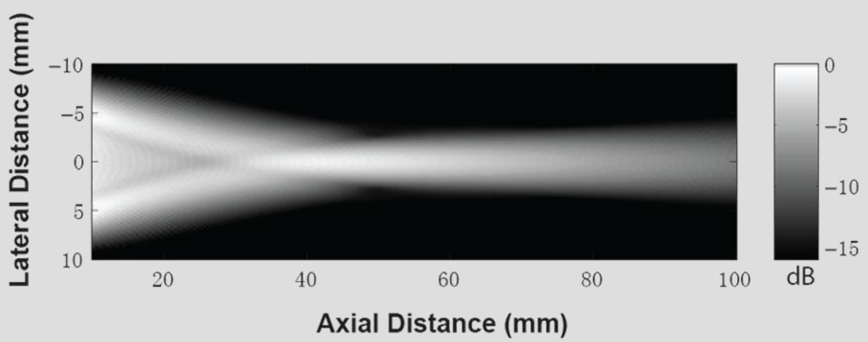


Fig. 3 Realistic beam profile generated by FieldII.

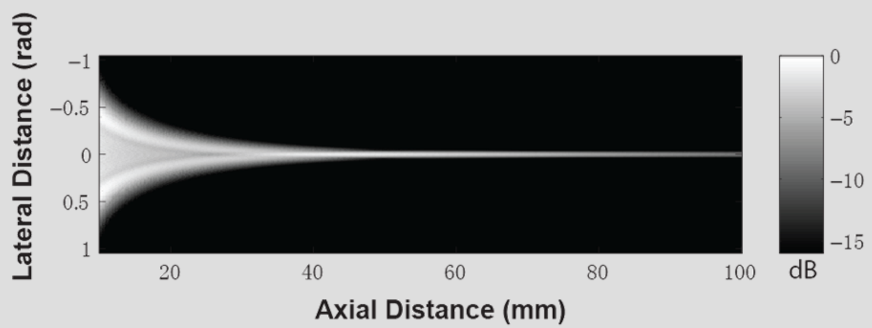


Fig. 4 LUT tabulated from realistic beam profile generated by FieldII.

Results

Lateral PSF: Narrow around the focus point

Amplitude of RF: Decreases with depth

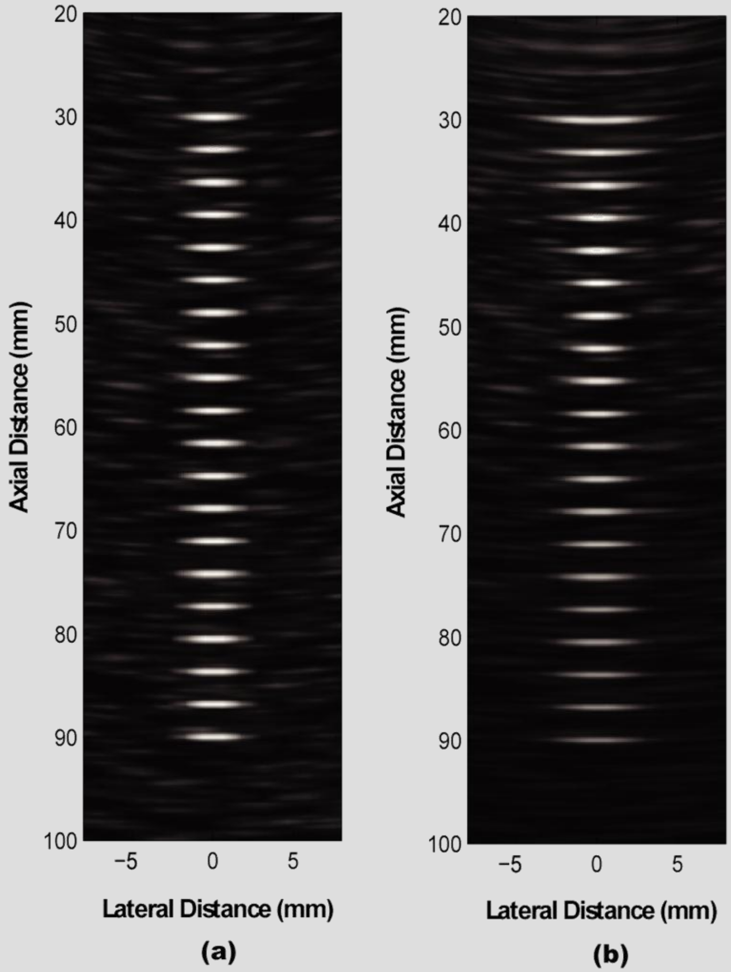


Fig. 5 The resulting images of COLE with analytical Gaussian PSF (a) and with realistic beam profile (b).

Conclusions

Computation efficiency:

LUT in Polar grids:

It peeds up computation by avoiding unnecessary (but time consuming) coordinate transformations

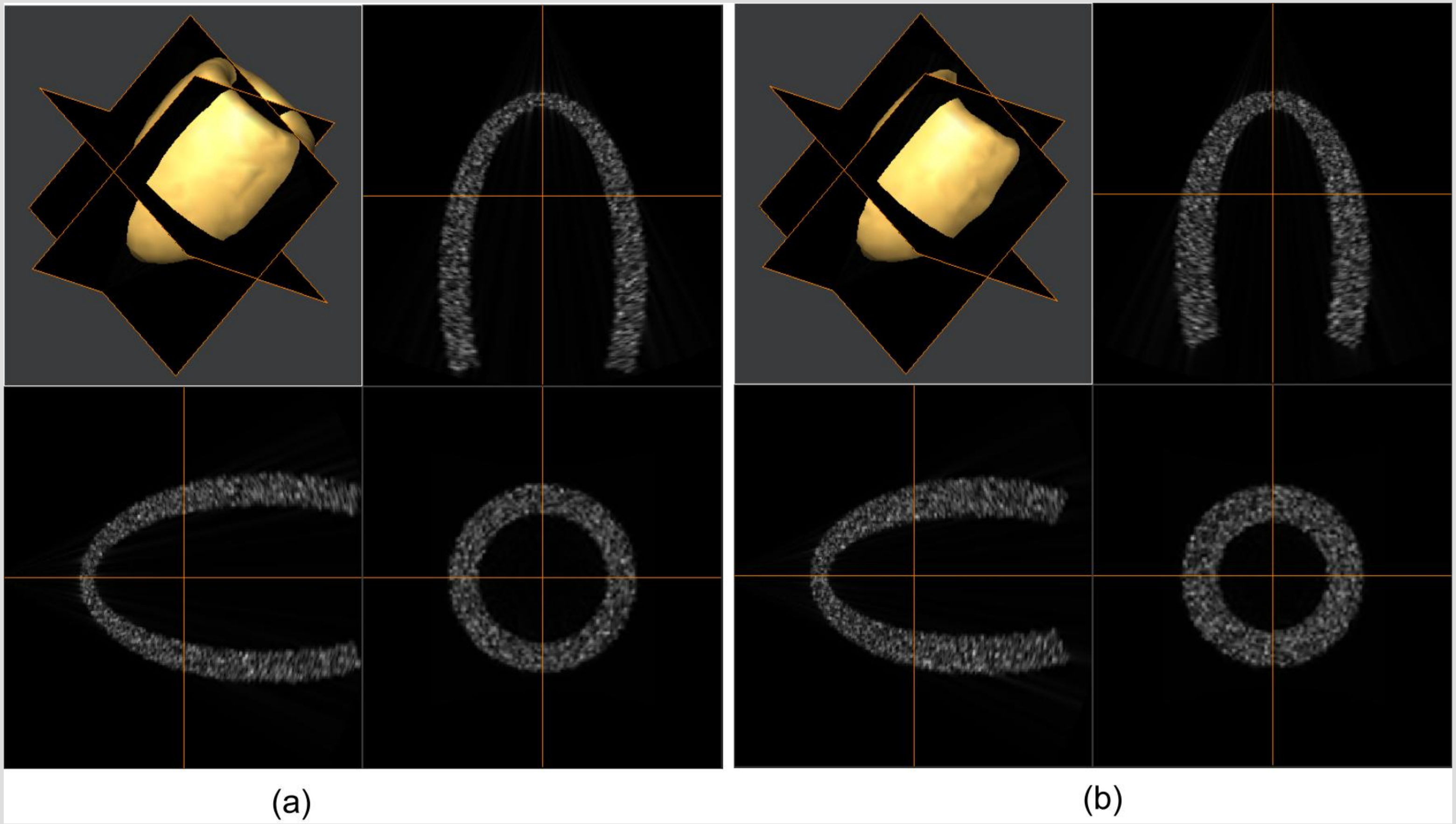
LUT for all image frames:

High computation efficiency will be obtianed in cardiac ultrasound where high frame rates are typically used – estimating the diffraction effects is thus only done once in COLE while it would be repeatedly estimated for each image frame using e.g. "FieldII"

A convolution-based methodology – COLE – was improved in order to integrate realistic beam profiles. This allows simulating cardiac images with more realistic image properties.

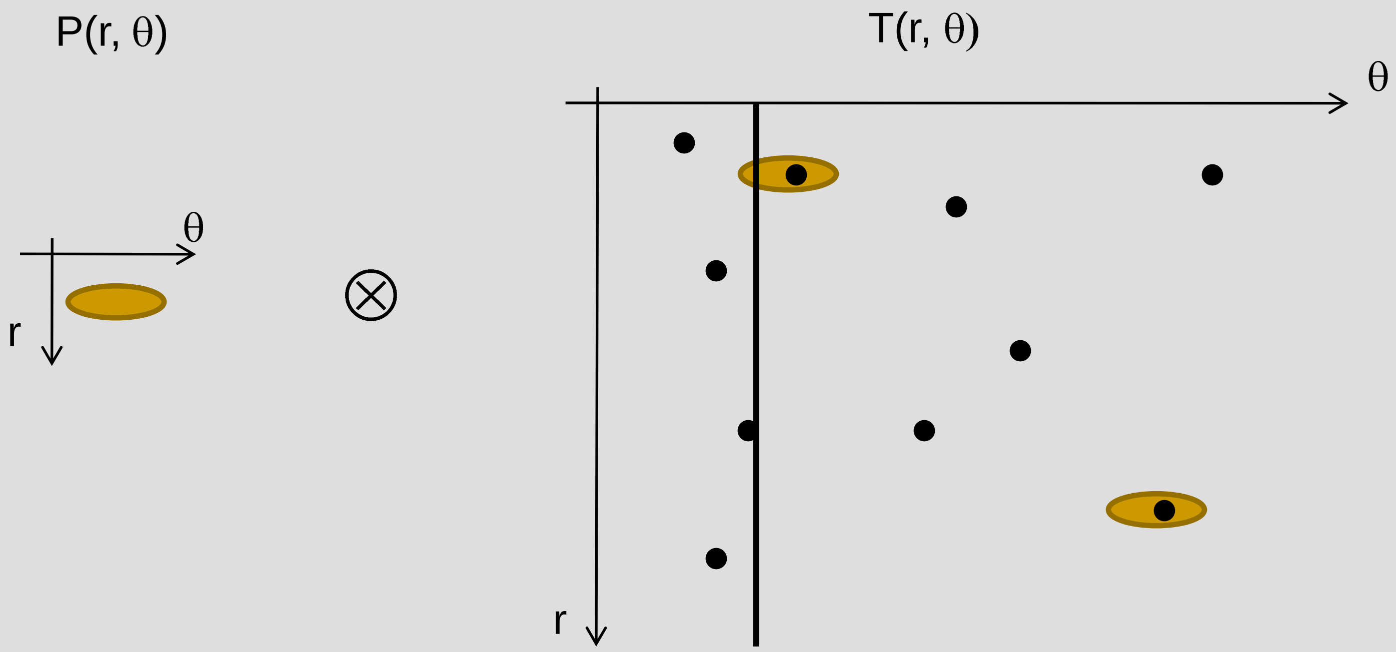
Simulation Results

3D US Simulation



The 3D surface of LV model based on the 3D US data sets of frame 1 during the movement of 60 bpm (a), and likewise the 3D surface of frame 10 during movement of 60 bpm (b).

COLE: graphically



Axial and lateral PSF are assumed to be separable

$$PSF(x,y) = PSF(x) \cdot PSF(y)$$