LA-UR-15-27376
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Title: Novel techniques for image quality enhancement in ultrasound imaging tomography

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Intended for: Job interview

Issued: 2015-09-22

Novel Techniques for Image Quality Enhancement in Ultrasound Imaging and Tomography

Jun Shin

Philips Research Seminar

September 23, 2015

Outline

I. **Image Quality Enhancement in Ultrasound Imaging**

   a. **Background**
   b. **Contrast Resolution: DAX & MPAX**
   c. **Signal-to-noise Ratio: FXPF**
   d. **Spatial Resolution: MVBF & Deconvolution**

II. **Image Quality Enhancement in Ultrasound Tomography**

   a. **Background**
   b. **Sound Speed Reconstruction**

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Ultrasound Imaging

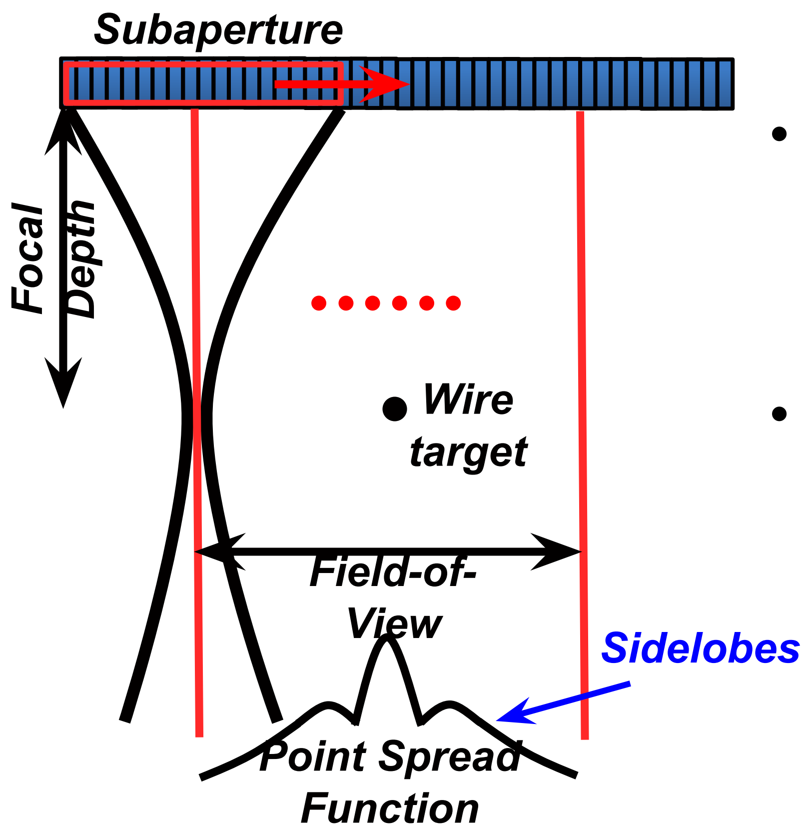
http://www.healthcare.philips.com

- **Major Advantages**
  - Non-invasive
  - No ionizing radiation
  - Portable
  - Inexpensive
  - Real-time

- **Challenges for High Image Quality**
  - **Poor Contrast Resolution**
    - Speckle noise, Off-axis clutter, Phase aberration, etc
  - **Limited Sensitivity**
    - Transmit power, Attenuation, Electronic noise, etc
  - **Limited Spatial Resolution**
    - Aperture size, Center frequency, Bandwidth, etc

Imaging with Linear Array

- **Delay-and-sum(DAS) beamforming**
  - Gold standard beamforming technique
  - Applies time delays based on path length differences
  - Coherent signals sum constructively while incoherent signals sum destructively

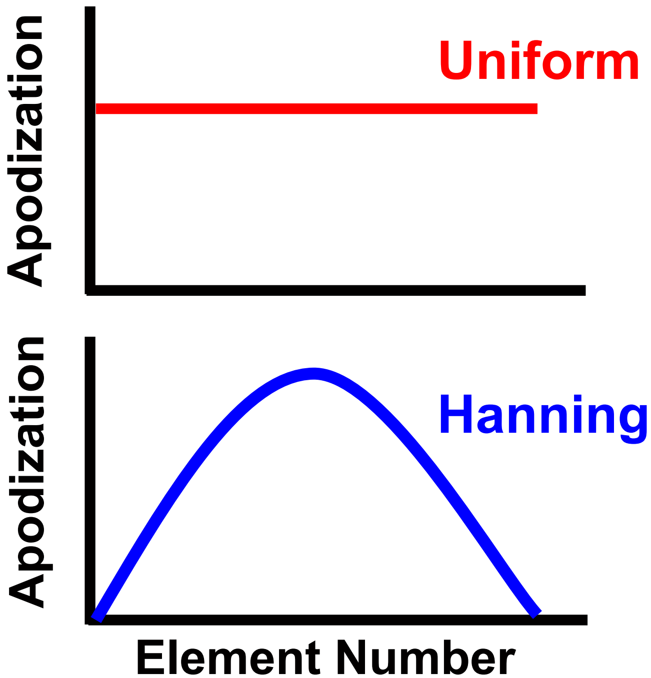


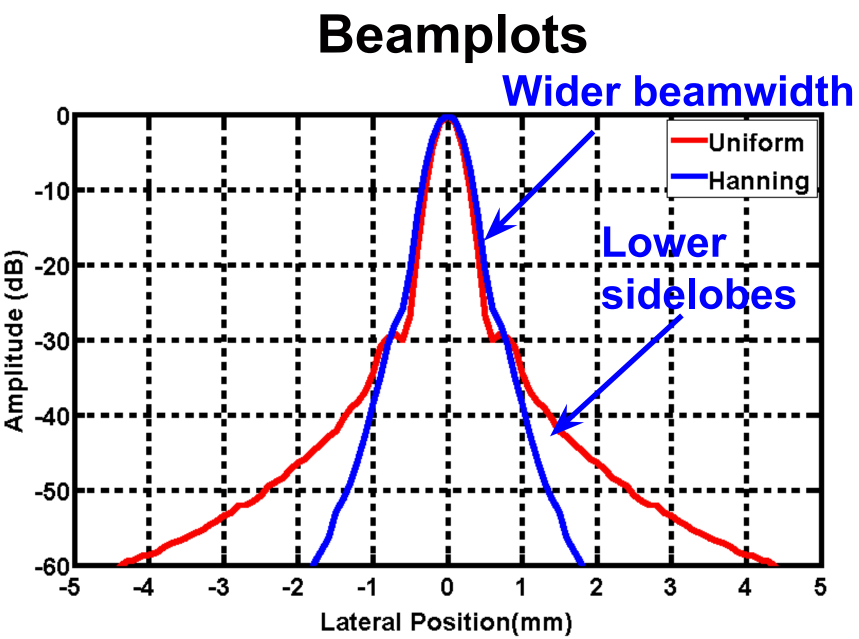
- **Infinitely long aperture**
  - can focus all energy at a single location

- **Finite aperture size**
  - Limited by anatomy
  - Hardware limitations
  - Off-axis sidelobes unavoidable

Apodization

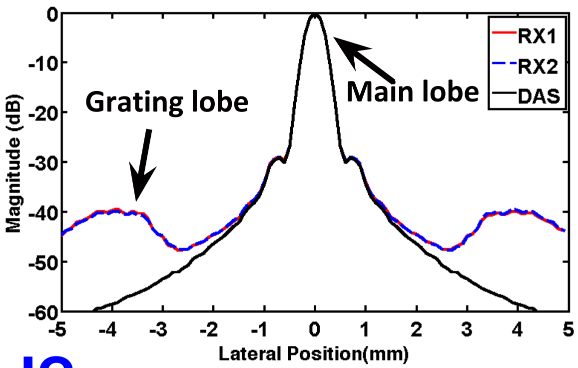
- **Weighting of channel RF signals across aperture**
  - Reduces sidelobes at the expense of broader mainlobe width





Grating Lobe: Friend or Foe?

- **Grating lobes:**
  - Mainlobe-like structures at certain angles from mainlobe



Foe?

- **Grating lobes are undesirable:**
  - Reduced contrast
  - Ghost images
- **Grating Lobe Reduction Techniques:**
  - Smaller pitch
  - Shorter TX pulses
  - Random aperture

Vs.

Friend?

Dual Apodization with Cross-correlation (DAX)

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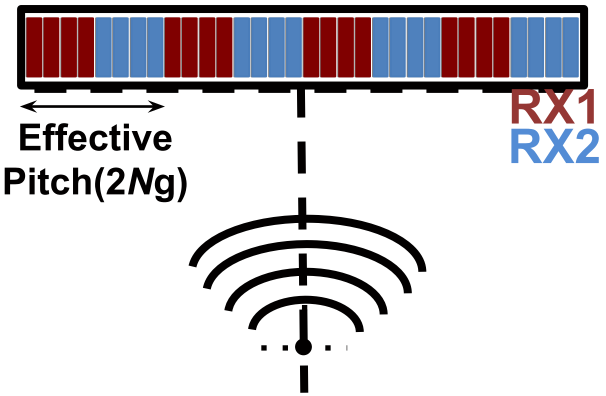
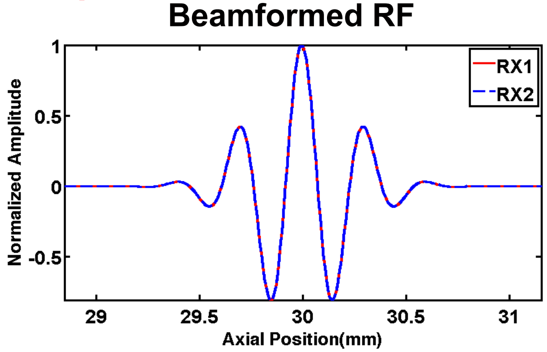
DAX: How it works

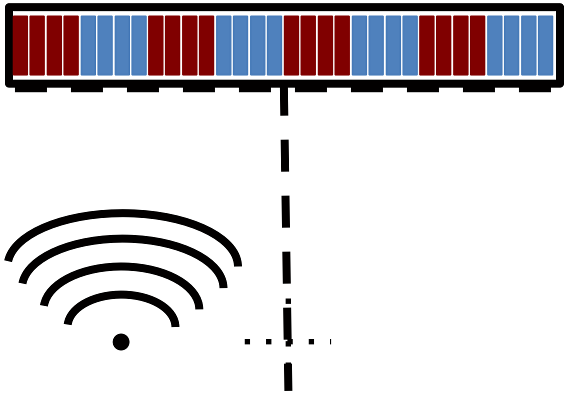
1. On-axis Target



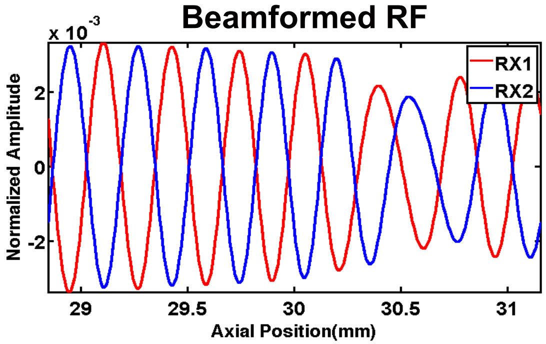
In phase!



2. Off-axis Target

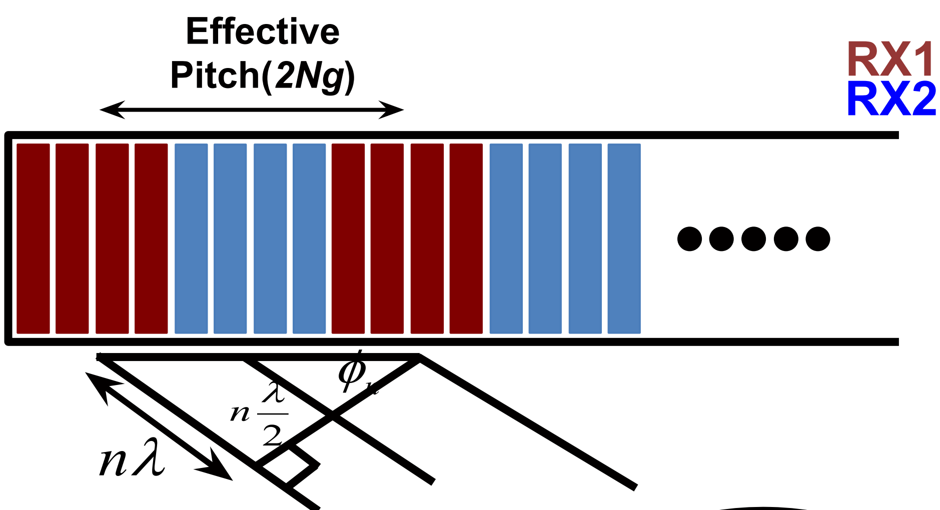


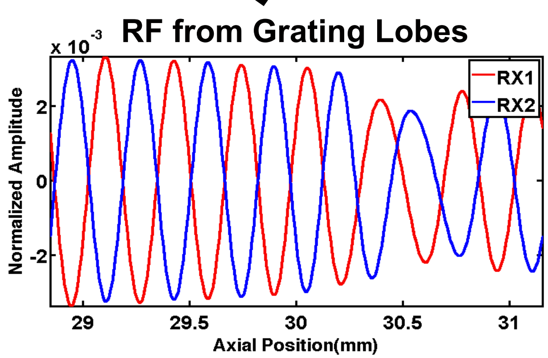
Out of phase!

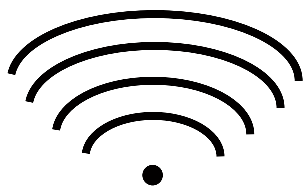


DAX: How it works

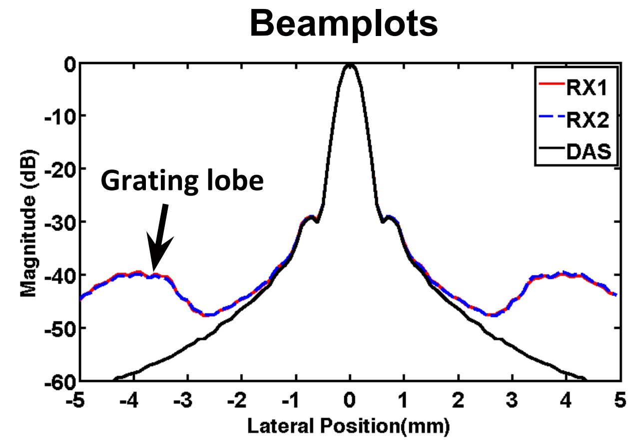
Grating Lobes







Target from grating lobe



Grating Lobe Angle

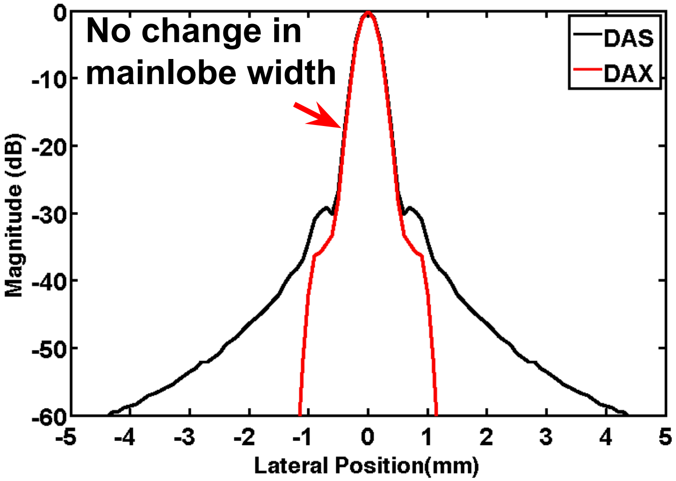
- **DAS:** $\phi_n = \sin^{-1}\left(\frac{n\lambda}{g}\right)$
- **N-N:** $\phi_n = \sin^{-1}\left(\frac{n\lambda}{2Ng}\right)$

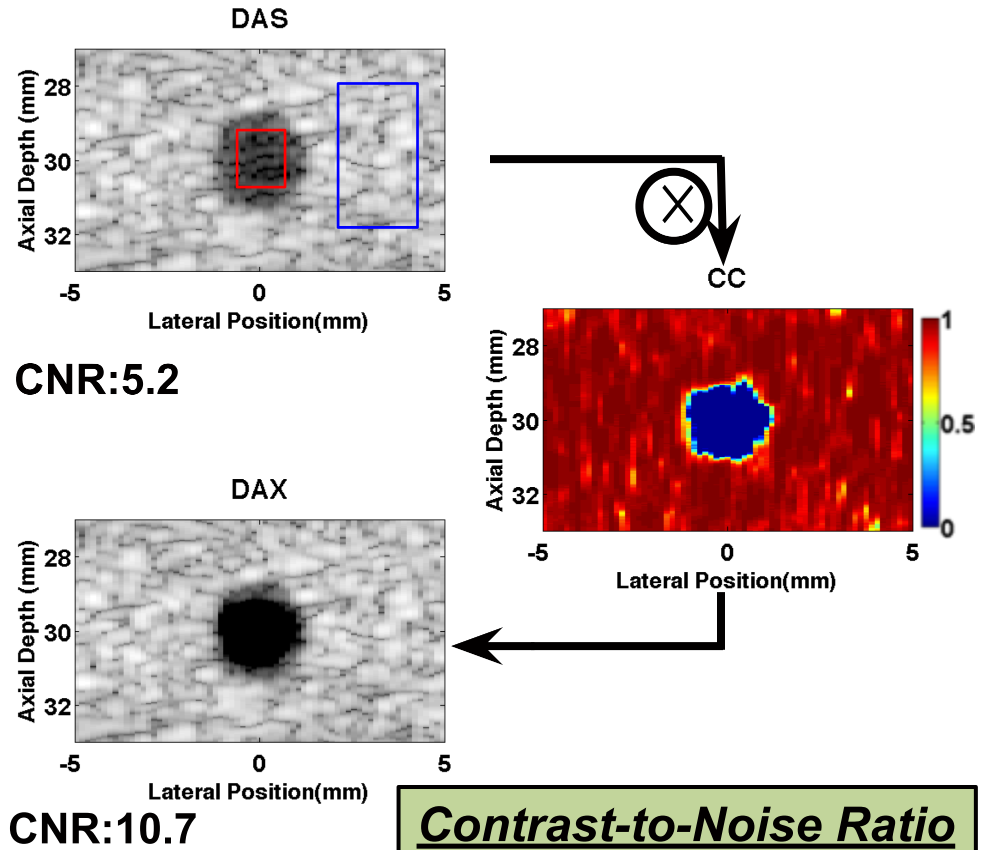
g = pitch
N = # of altern. elems

180° phase difference between RX1 and RX2 → **Normalized Cross-correlation**

DAX: Simulation Examples

With an 8-8 Alternating Pattern:





- Large suppression of sidelobes
- No trade-off in lateral resolution!

Contrast-to-Noise Ratio

$$CNR = \frac{\left| \overline{S_t} - \overline{S_b} \right|}{\sigma_b}$$

DAX: Advantages & Limitations

Advantages

- Straightforward
- Computationally cheap
- Large CNR improvement
- No loss in spatial/temporal resolution
- Robust with weak-medium level aberrations

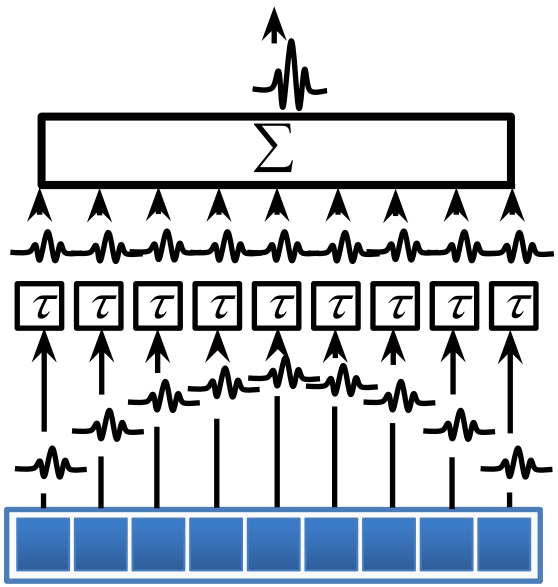
Limitations

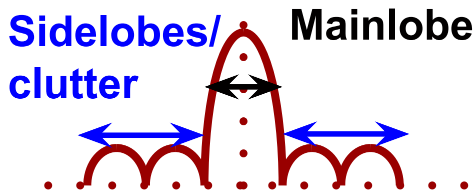
- Creates artifacts & reduced effectiveness with strong phase aberrations
- Limited performance in the presence of reverberation clutter

Possible Solutions

- Integrate with other imaging techniques
- Algorithm Modifications / Optimizations

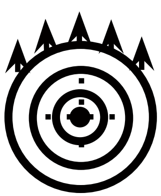
Problem I: Phase Aberration





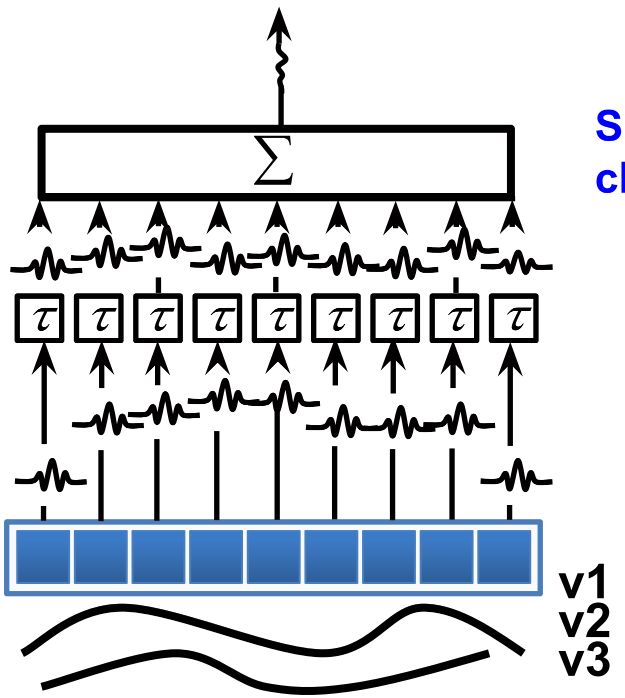
No errors

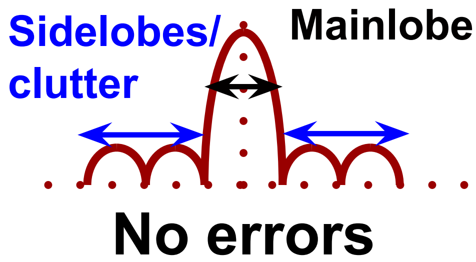
- Best spatial resolution
- Largest target amplitude



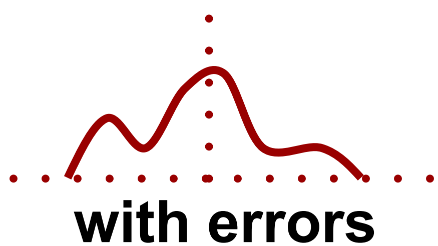
C = 1540 m/s

Problem I: Phase Aberration

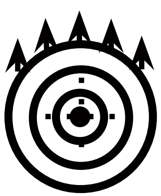




- Best spatial resolution
- Largest target amplitude



- Wider main lobe
- Higher sidelobes
- Lower target amplitude

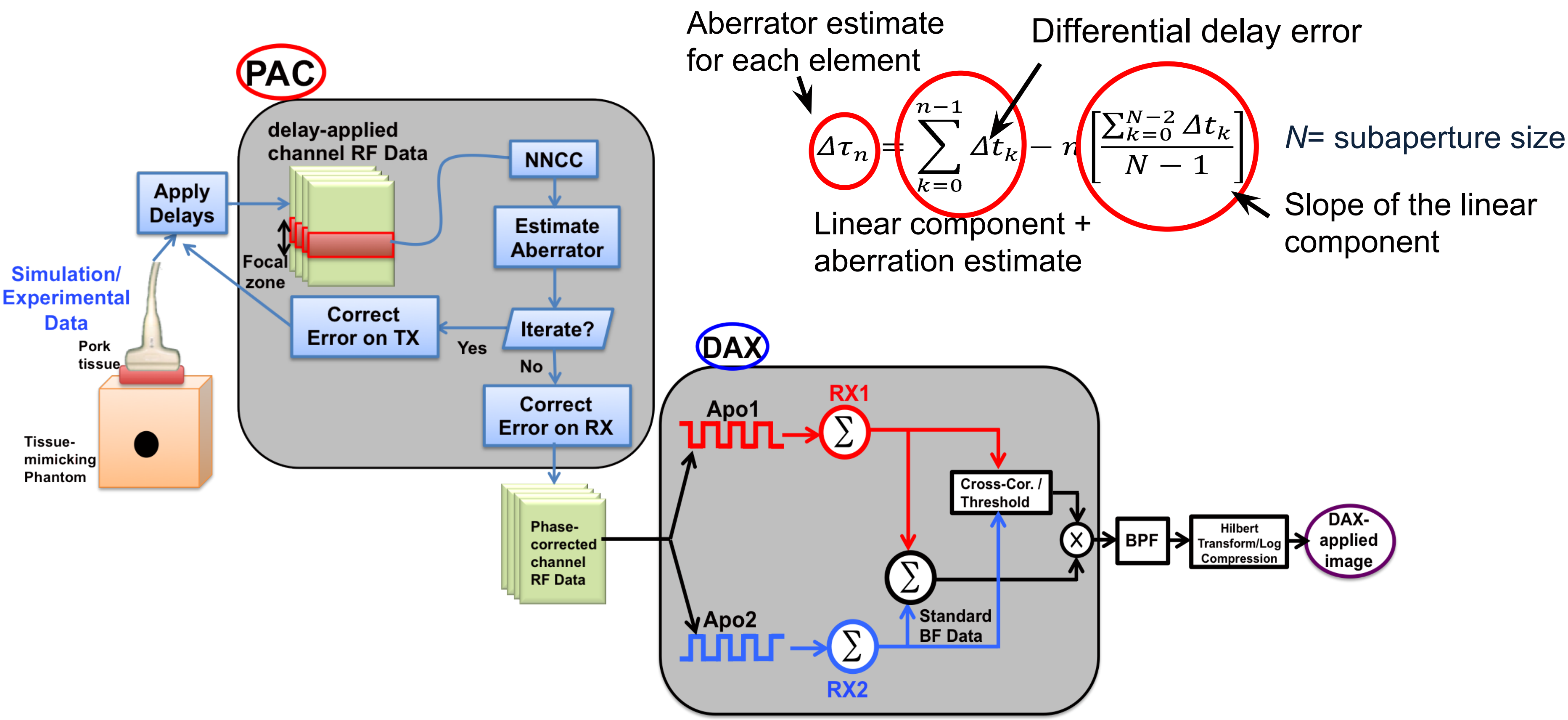


C ≠ 1540 m/s

Soft tissue sound speed range:
1350 m/s – 1725 m/s

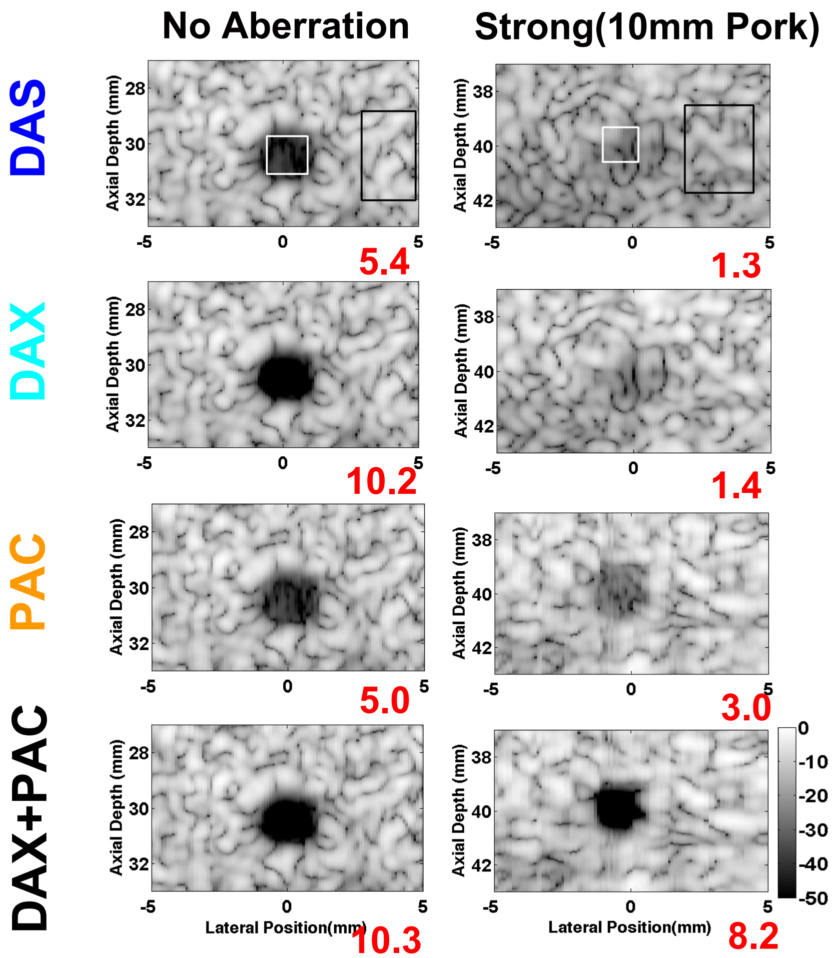
Solution I: DAX + PAC

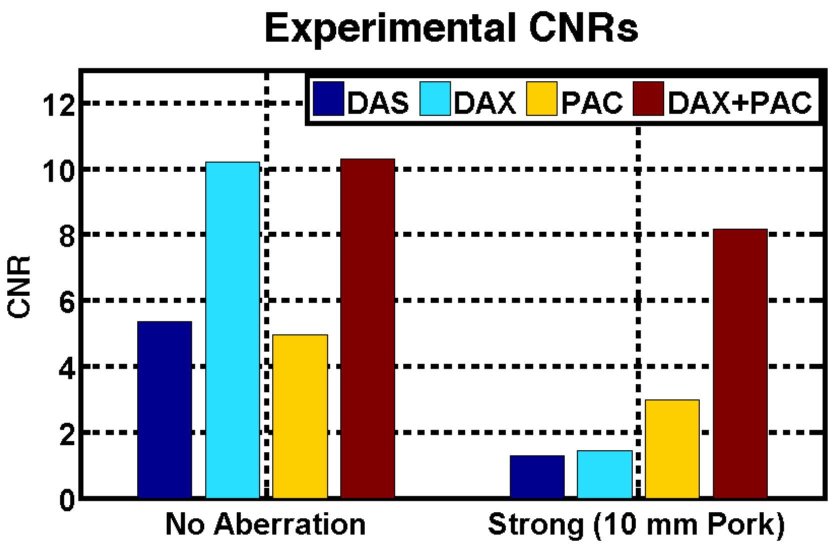
- **Phase Aberration Correction (PAC) + DAX**



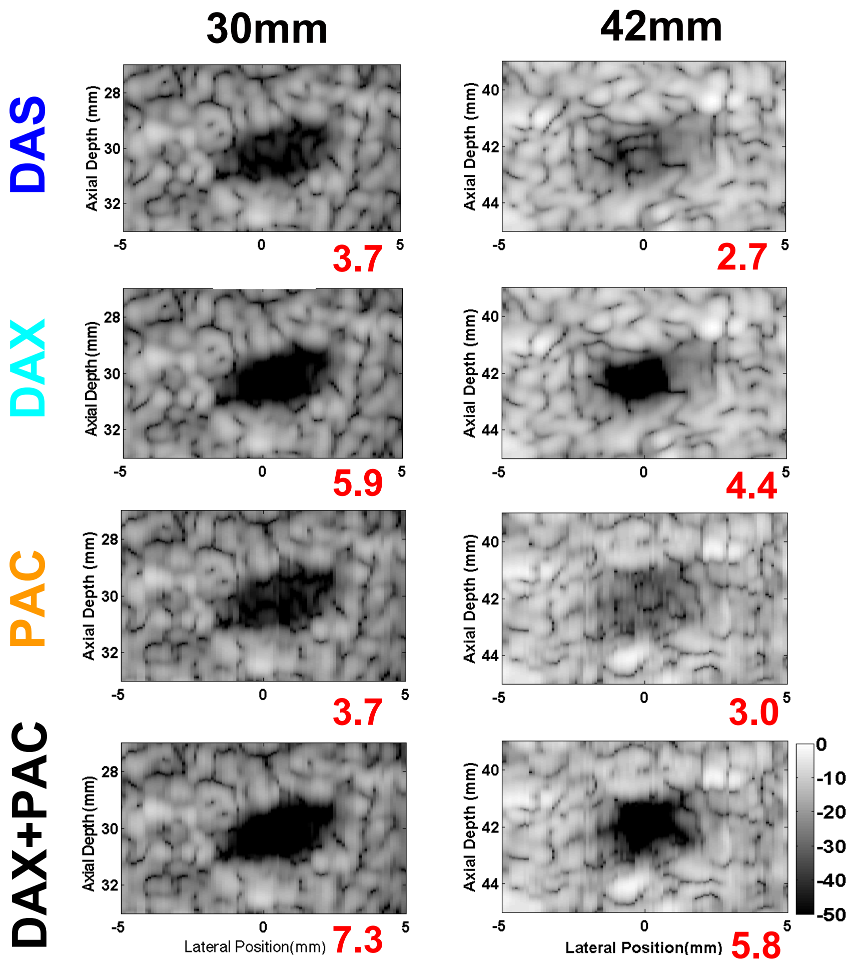
- **Benefit from 2 independent contrast enhancement mechanisms**
  - PAC restores coherence lost due to aberration (1 iteration)
  - DAX suppresses the remaining aberration effects

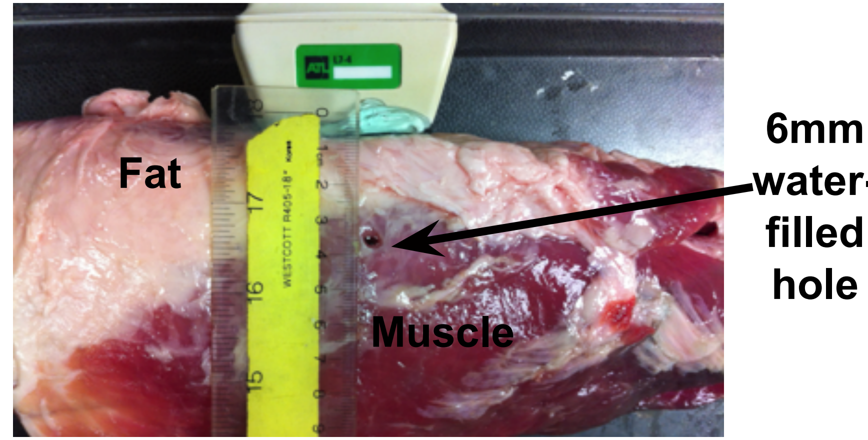
DAX + PAC: Experimental Cyst Results

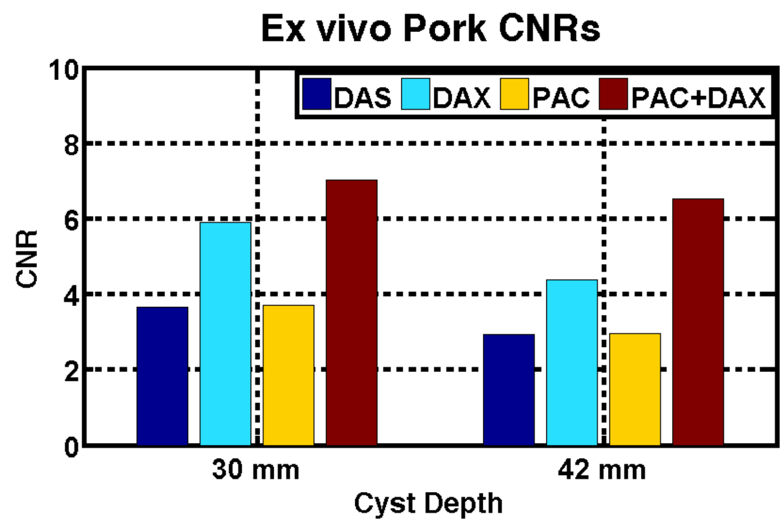




DAX + PAC: Ex-vivo Pork Results



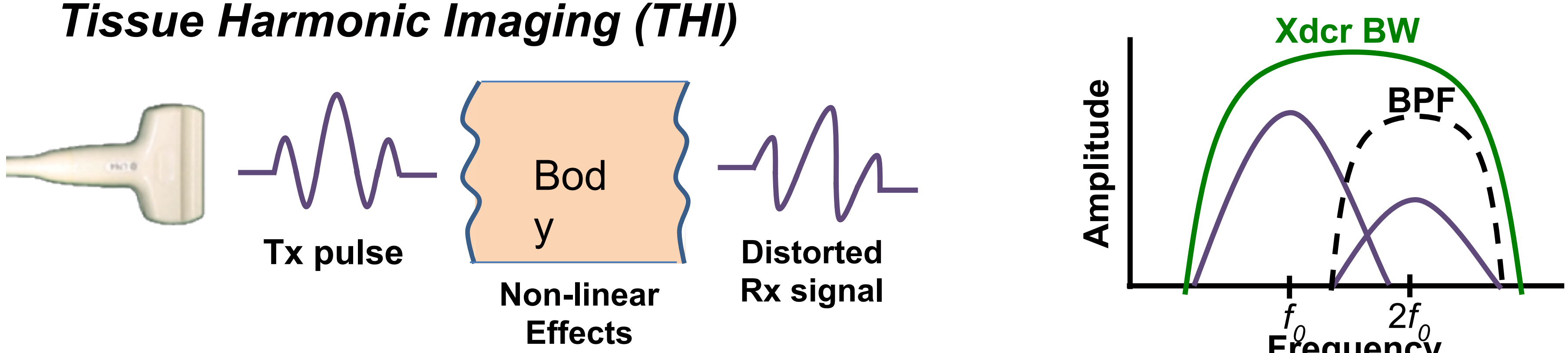




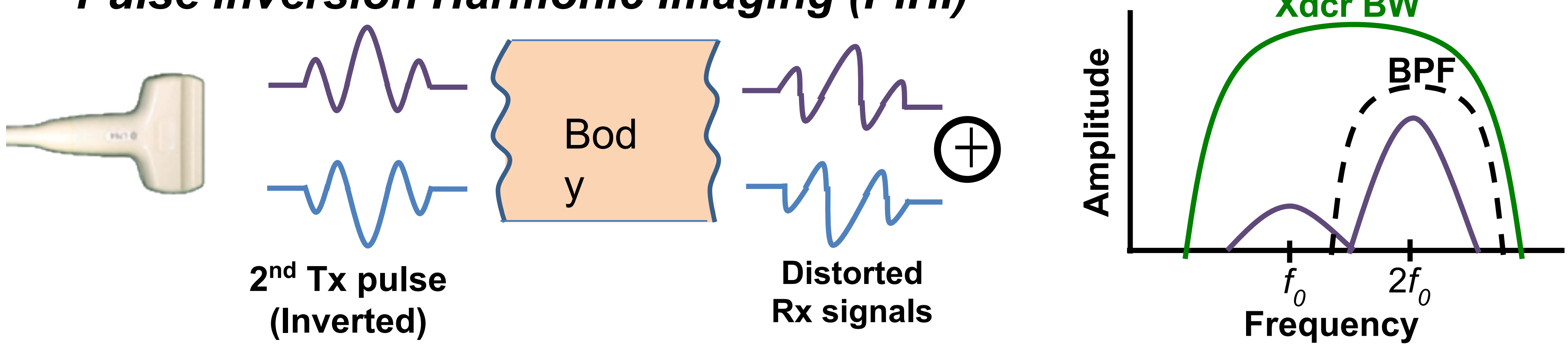
Solution II: DAX + HI

- **Harmonic Imaging (HI)**
  - One of the most important recent innovations
  - Default mode for many clinical applications (esp. cardiac)
  - Due to nonlinear effects of body tissue

Tissue Harmonic Imaging (THI)



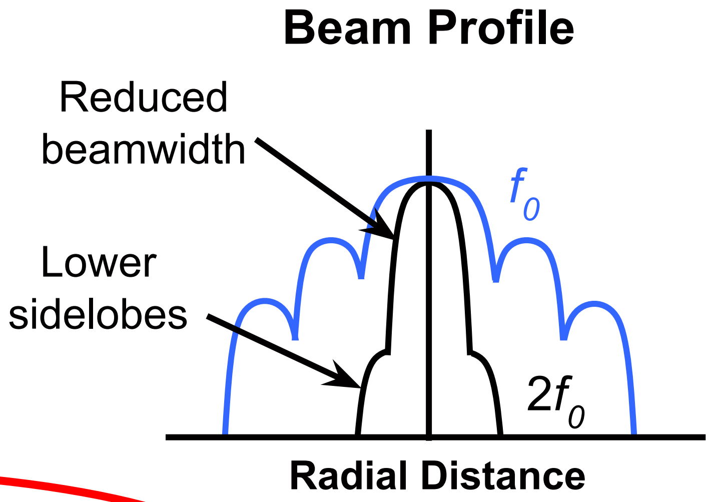
Pulse Inversion Harmonic Imaging (PIHI)



Why Harmonic Imaging?

Advantages

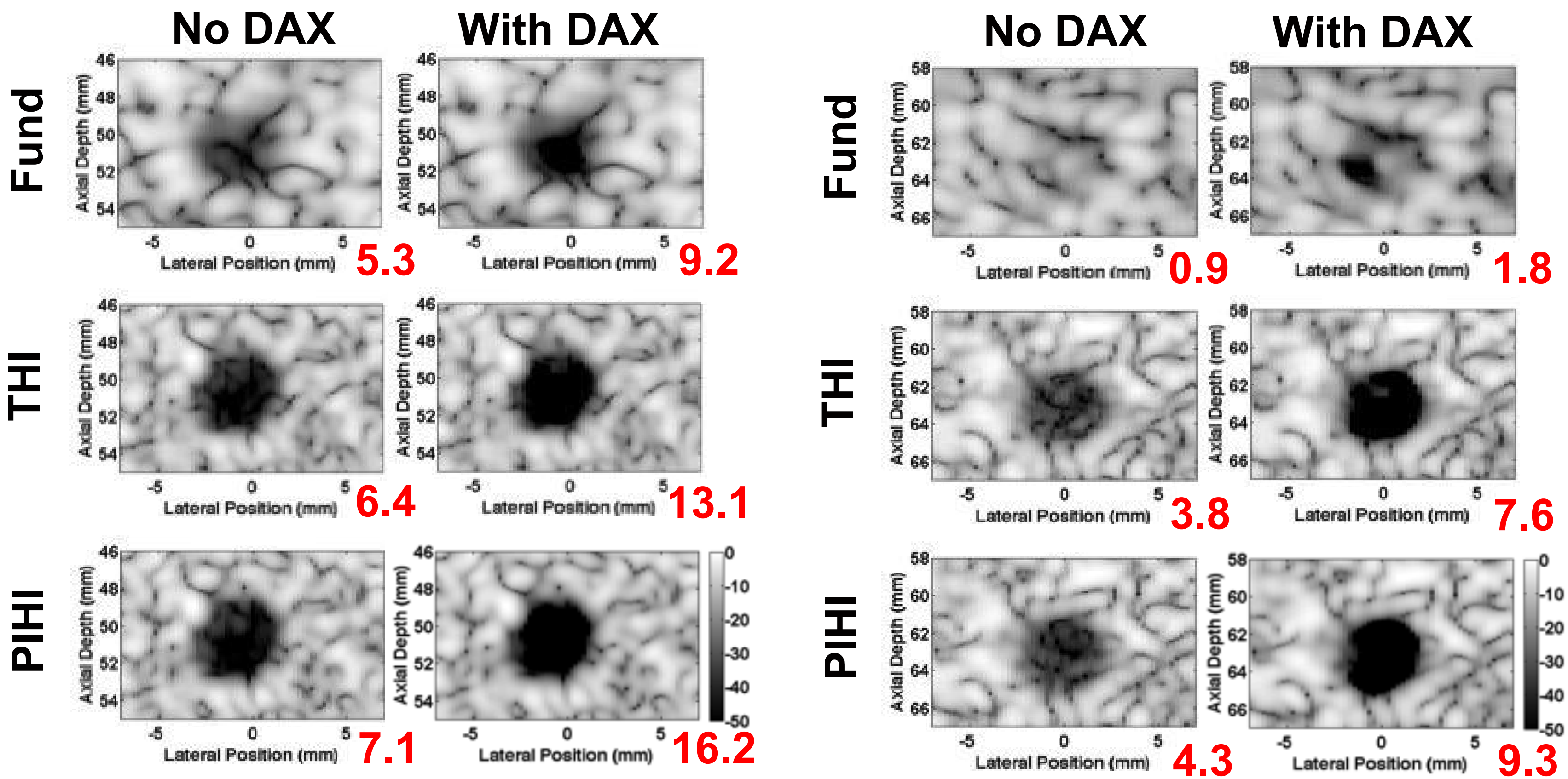
- Sharper image with higher contrast
- Most acoustic noise dominated at f_0
- Aberration effects suppressed
- No added computational burden



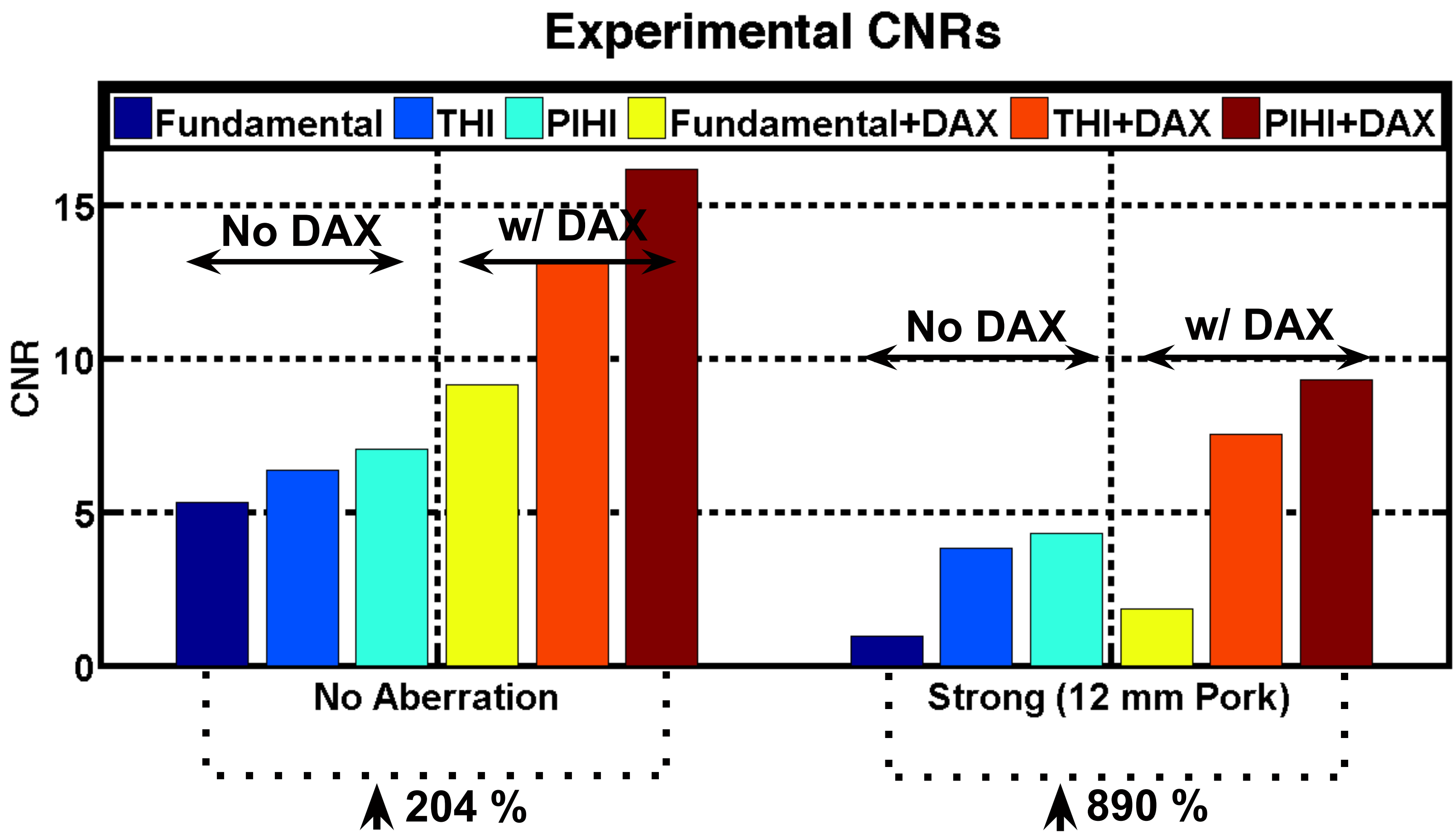
DAX can benefit from this!

DAX + HI: Experimental Cyst Results

No Aberrator | **12 mm Pork Aberrator**



DAX + HI: CNR Summary



Summary: Solutions to Phase Aberration Effects

- **Solution I: DAX + PAC**

	% CNR Improvement		Image Artifacts	TX Firings
	Phantom Experiment	**Ex-vivo Pork Experiment**		
DAX	13 %	64.9 %	Yes	1
PAC	135 %	11.3 %	No	2
DAX+PAC	**543 %**	**117 %**	**No**	**2**

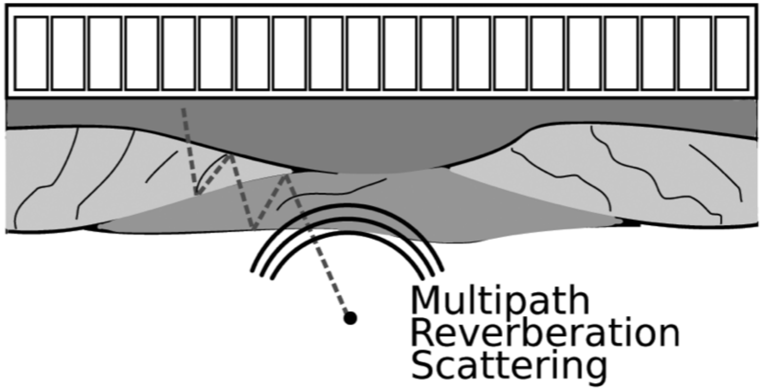
- **Solution II: DAX + HI**

	% CNR Improvement (Phantom Experiment)		Image Artifacts
	No Aberrator	**12-mm Pork**	
Fund+DAX	72 %	96 %	Yes
PIHI	33 %	359 %	No
PIHI+DAX	**204 %**	**890 %**	**No**

Problem II: Reverberation Clutter

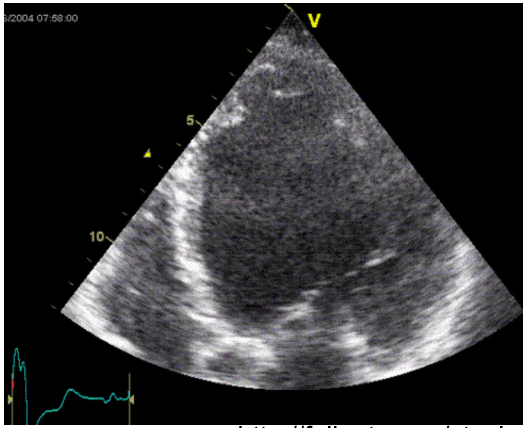
Reverberation

- 1 of the 2 primary sources of ultrasound image degradation (Aberration & Reverberation)
- Caused by near-field structures (tissue layers, ribs, etc)
- Dominant mechanism of image quality degradation for B-mode



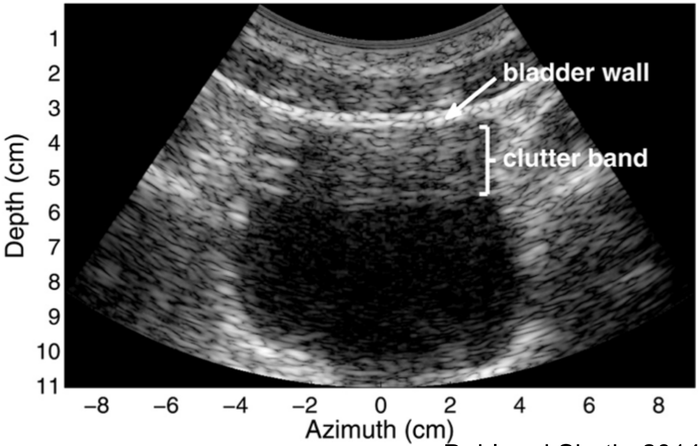
Byram and Jakovljevic, 2014

Cardiac Apical 4-Chamber View



http://folk.ntnu.no/stoylen/

Human Bladder



Dahl and Sheth, 2014

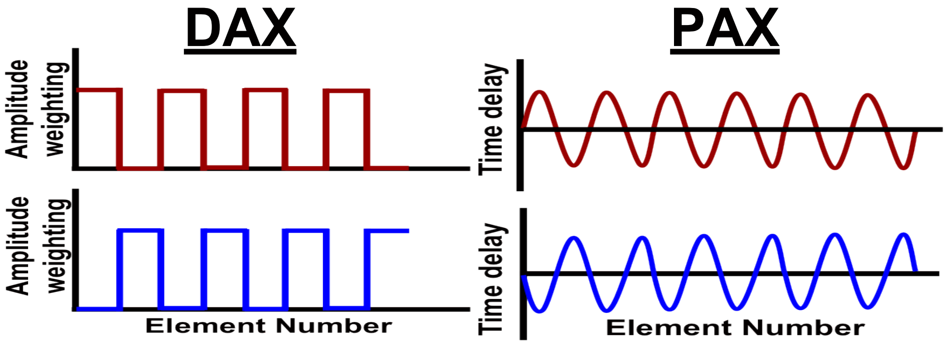
DAX grating lobes < -30 dB
Reverb clutter reduction is key for *in-vivo* imaging!

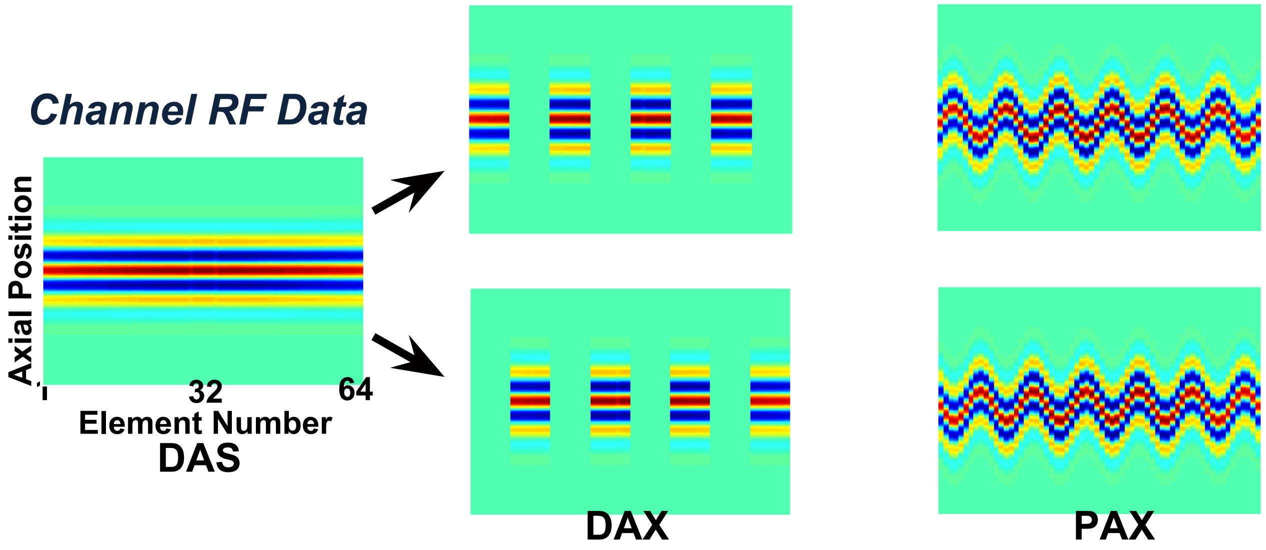
DAX vs. PAX

<u>PAX:</u> ***"Phase" Apodization with Cross-correlation***

- Motivated by the concept of sinusoidal phase grating in Fourier Optics (Goodman, 2005)

- **<u>DAX:</u>** Two sets of RF data with complementary amplitude weightings
- **<u>PAX:</u>** Two sets of RF data with complementary sinusoidal time delays





PAX: Overview

- Rayleigh-Sommerfeld Diffraction Theory
  - Describes complex pressure field at single frequency

Transmit Field

Receive Field

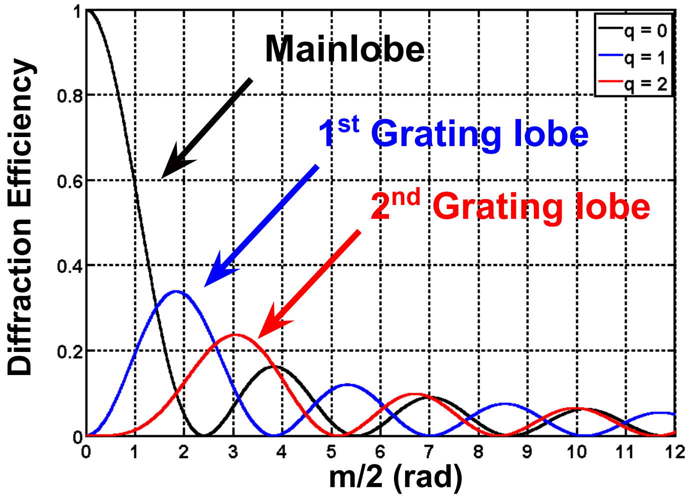
Sinusoidal phase apodization

$$= \sum_{q=-\infty}^{\infty} J_q\left(\frac{m}{2}\right) e^{j2\pi q f_0 x_0}$$

$$\psi_\omega(x,z) \approx A_t(u_x) \int_{-\infty}^{\infty} a_r(x_0) e^{-\frac{j2\pi x x_0}{\lambda z}} e^{j\frac{m}{2}\sin(2\pi f_0 x_0)} dx_0$$

$$\approx A_t(u_x) \sum_{q=-\infty}^{\infty} J_q\left(\frac{m}{2}\right) A_r(u_x - q f_0)$$

Bessel function of 1st kind



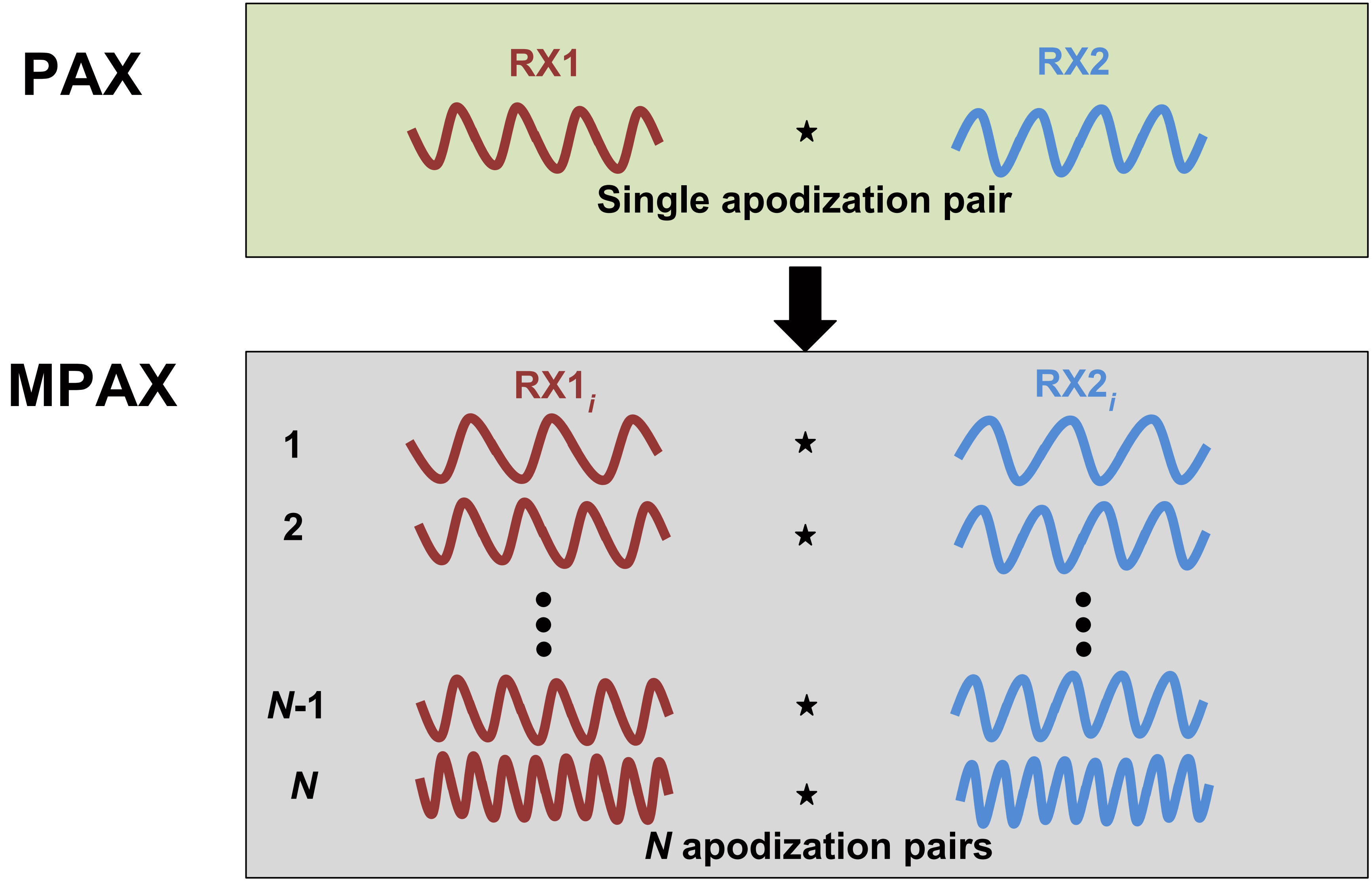
m : determines grating lobe magnitude

$\mathbf{f_0}$: determines grating lobe location

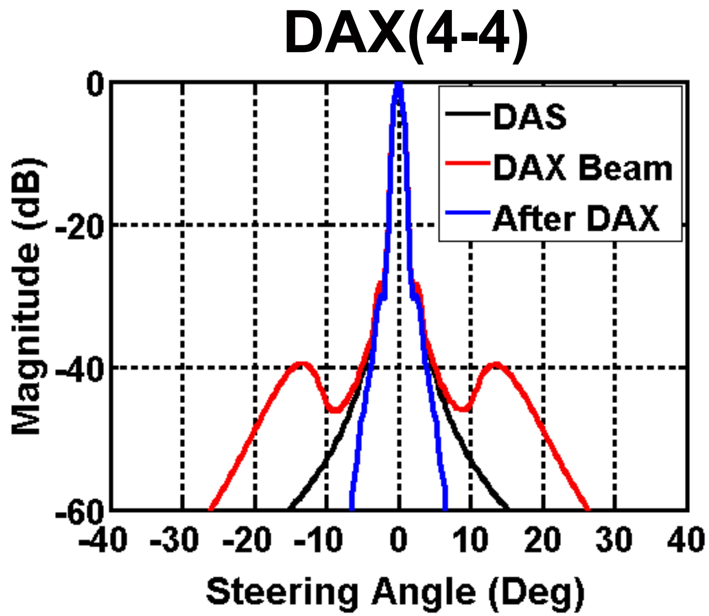
Sinusoidal phase apodization

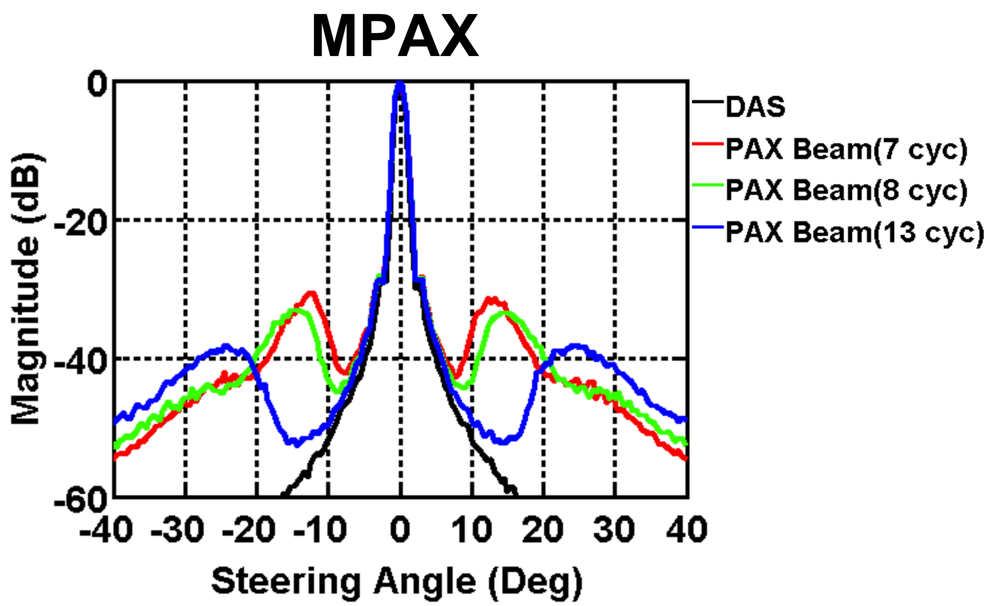
- Deflects main beam energy out to multiple grating lobes
- Allows for more flexible manipulation of grating lobe locations & magnitudes

From PAX to Multi-PAX (MPAX)



Beamplots: DAX vs. MPAX



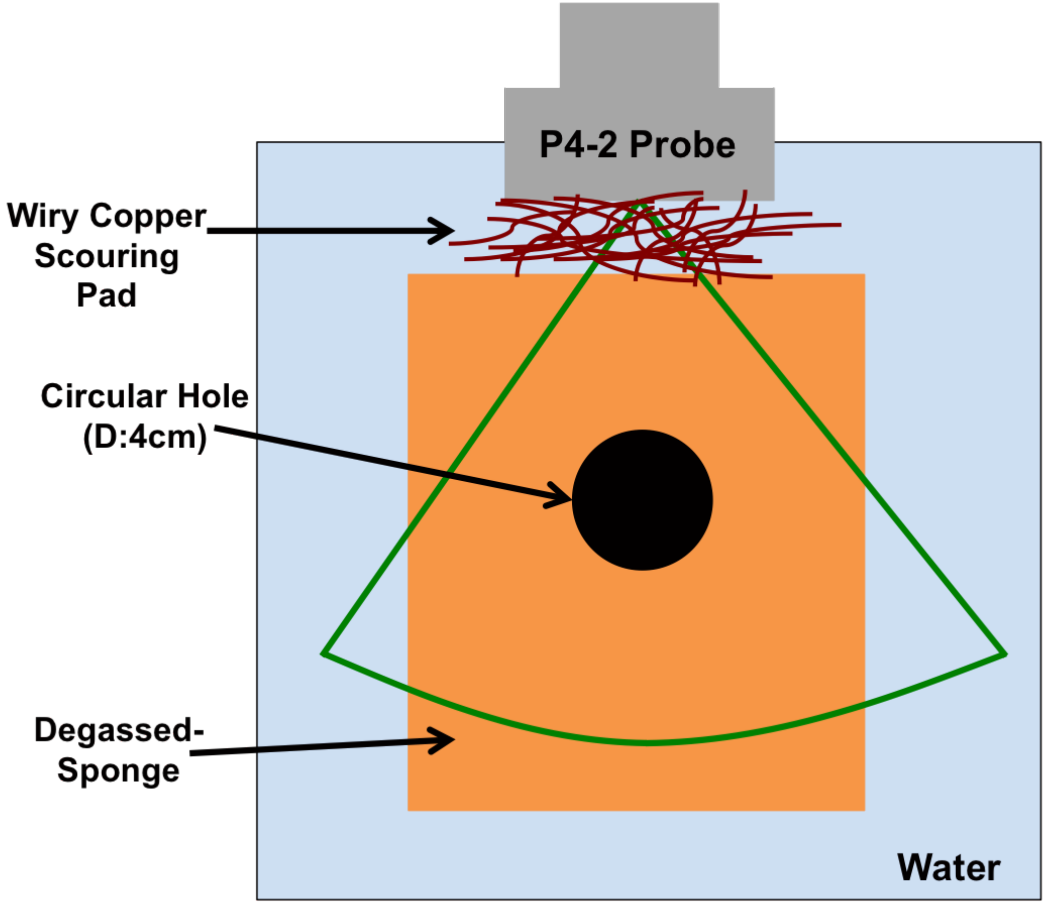


<u>MPAX</u> **: uses multiple PAX beams with 7~13 cycles (only 3 are shown)**

- ***m*** **= 3.6 rad (~ 0.6λ)**

➢ **MPAX uses the average of multiple coefficients for more robust performance.**

Sponge Phantom Experiment



Wiry Copper Scouring Pad

Copper wire:

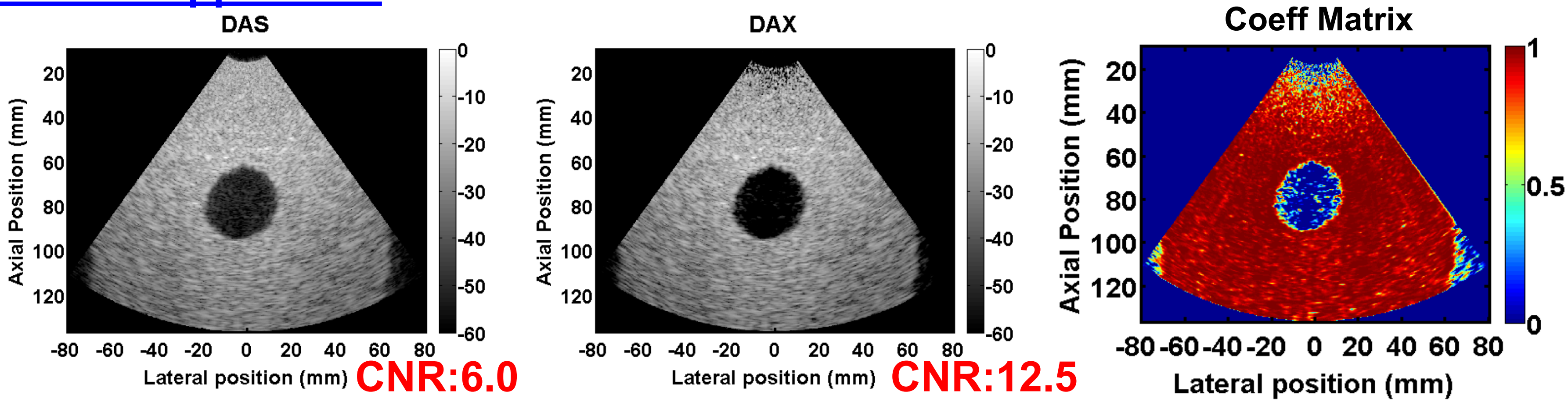
- Generates near-field reverb clutter similar to *in vivo*

Sponge:

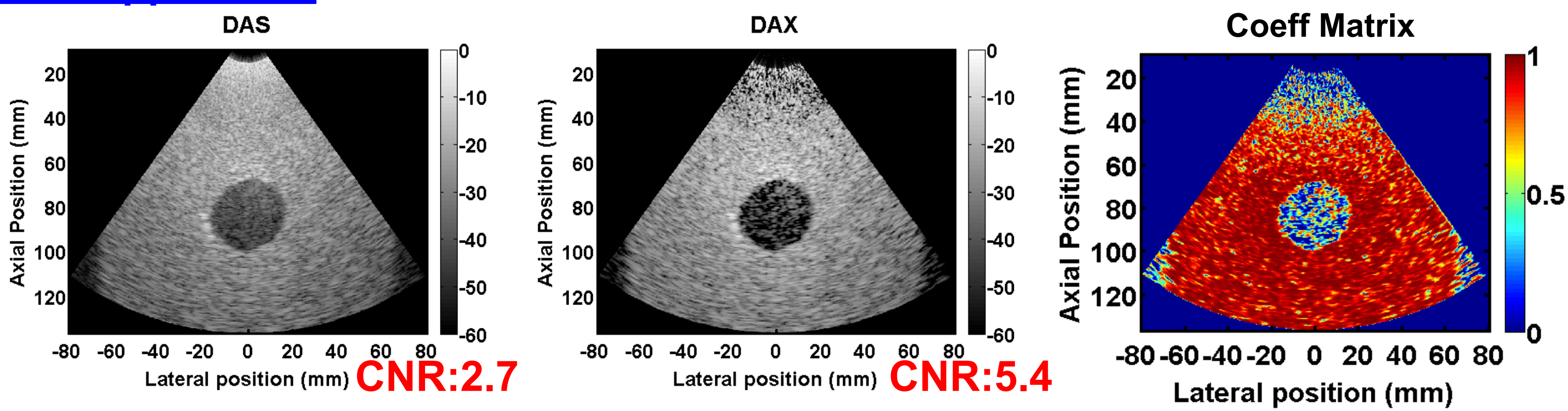
- Speckle-generating target

Sponge Results: DAX

Without Copper Wire

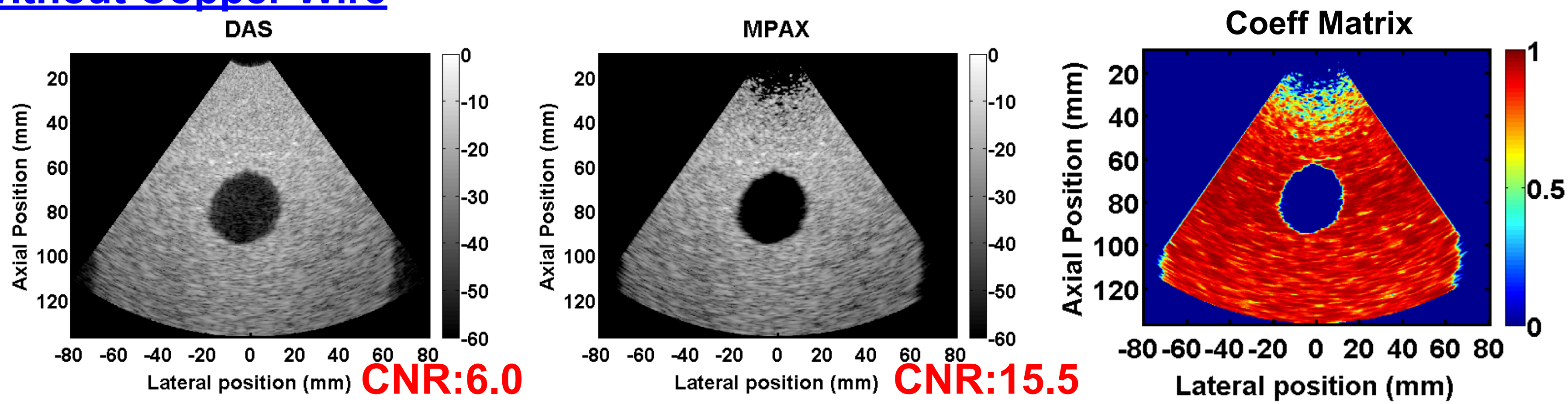


With Copper Wire



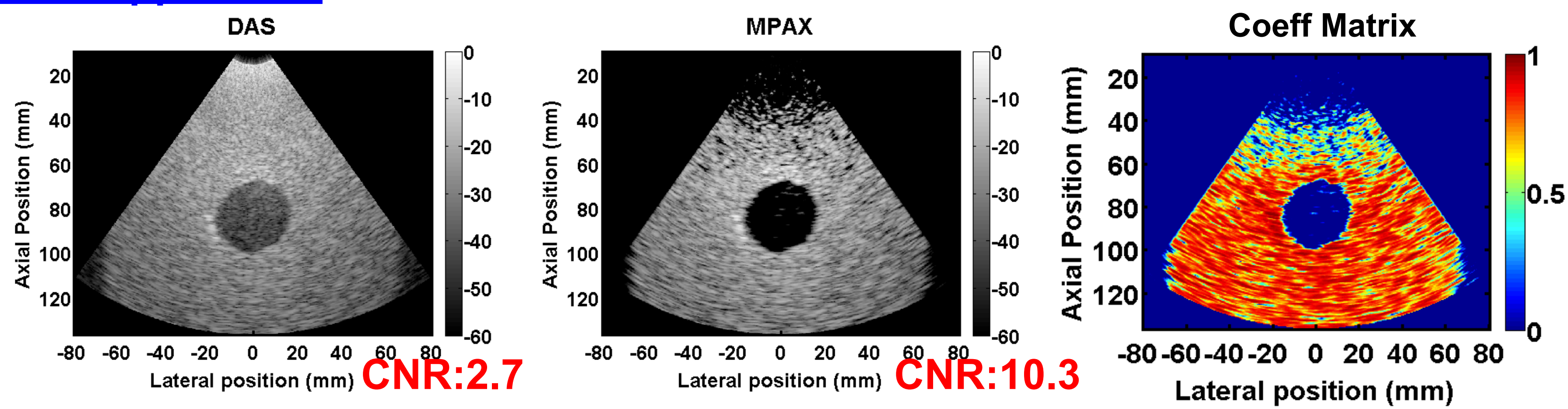
Sponge Results: MPAX

Without Copper Wire



CNR:6.0 CNR:15.5

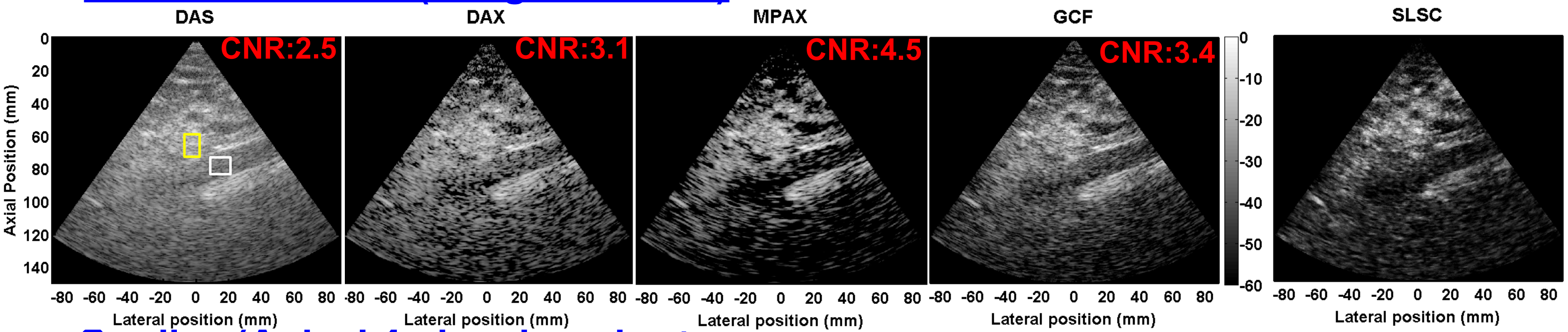
With Copper Wire



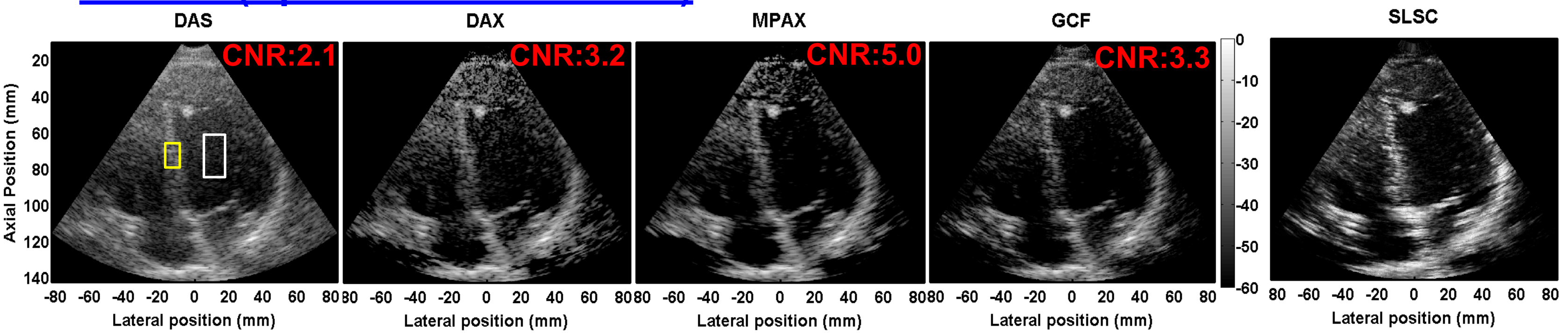
CNR:2.7 CNR:10.3

In vivo Evaluation

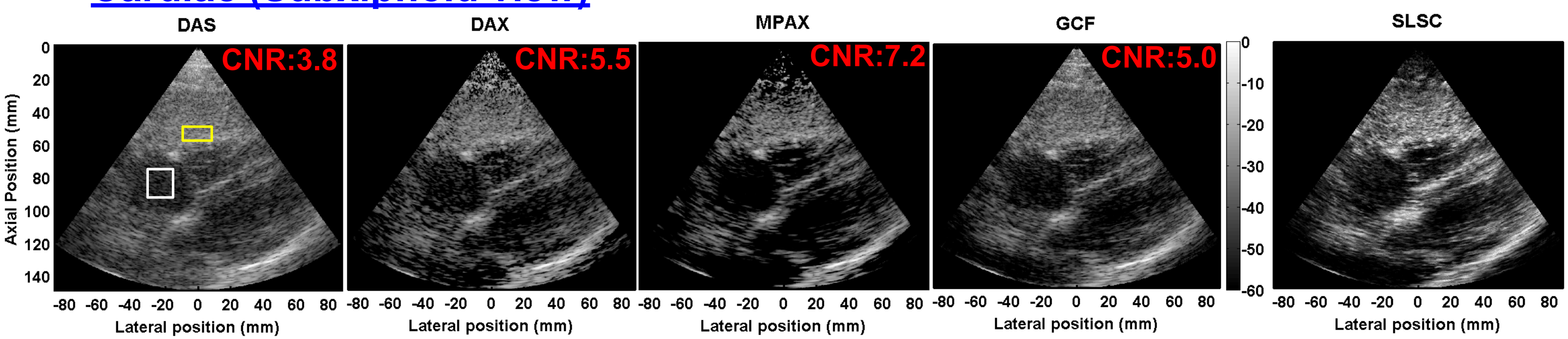
Abdominal Aorta (Long-axis view)



Cardiac (Apical 4-chamber view)

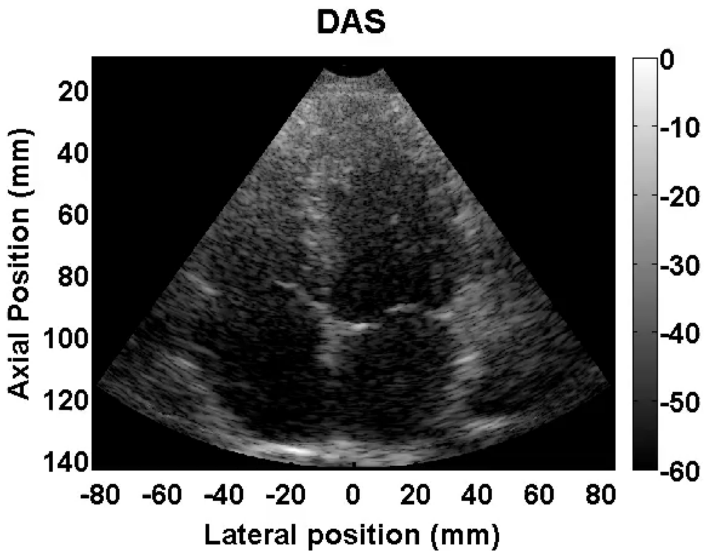


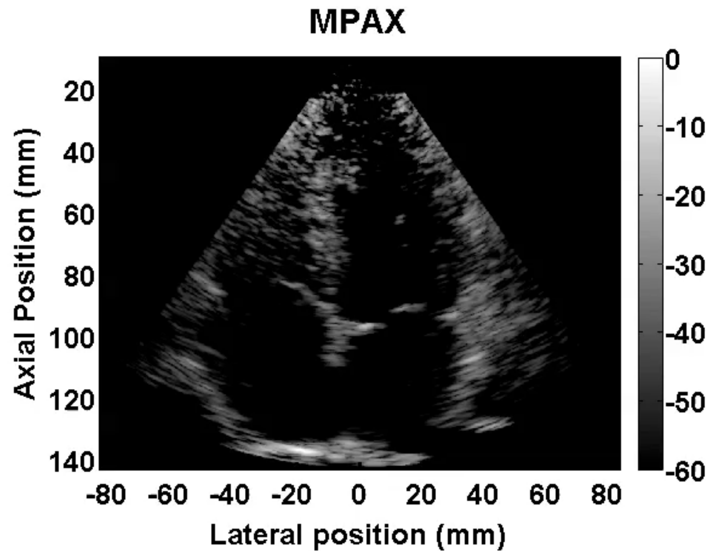
Cardiac (Subxiphoid view)



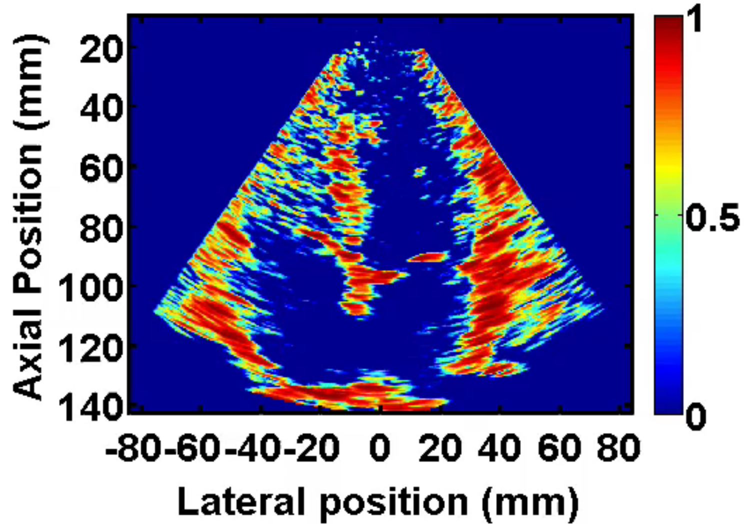
Cardiac Imaging: *In vivo* Evaluation

Apical 4-chamber View



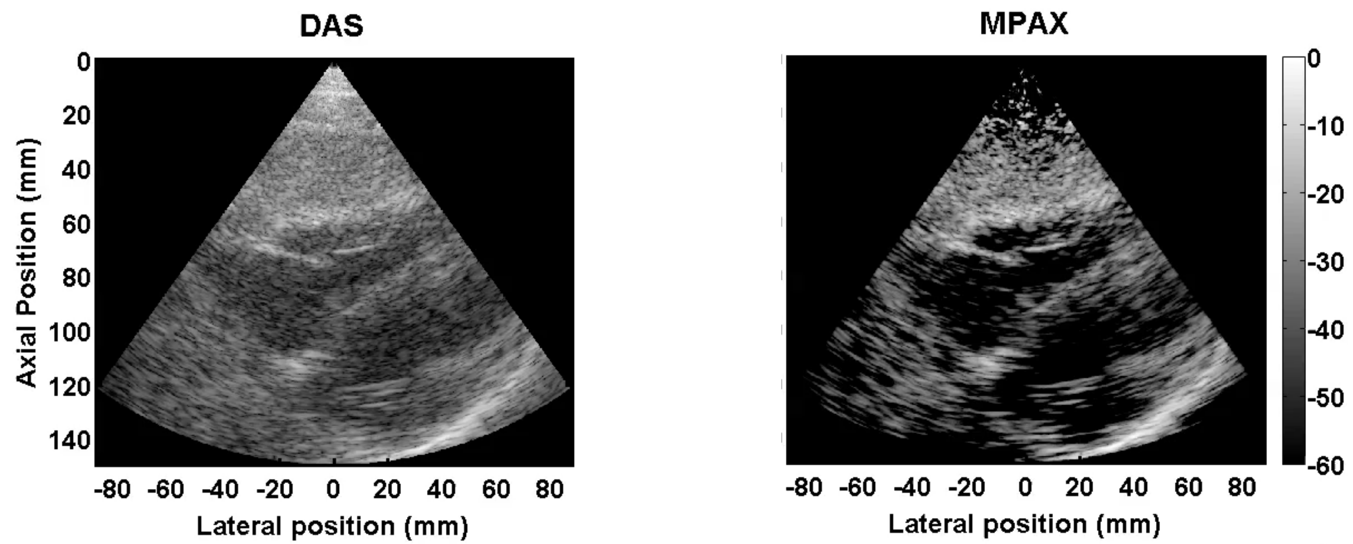


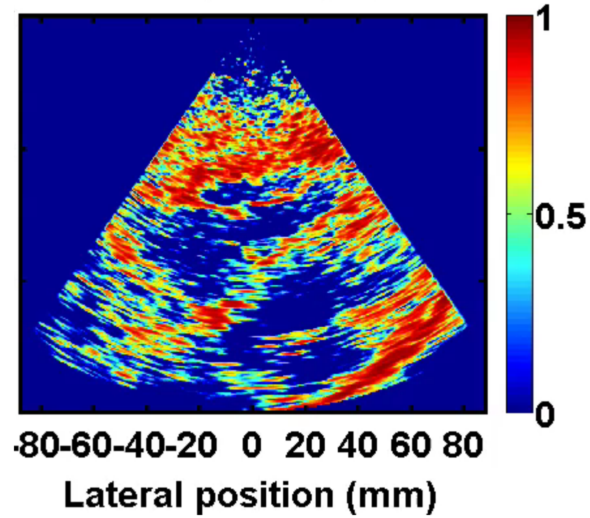
		DAS	MPAX
CNR	End-systole	2.1	6.7
	End-diastole	2.1	5.0



Cardiac Imaging: *In vivo* Evaluation

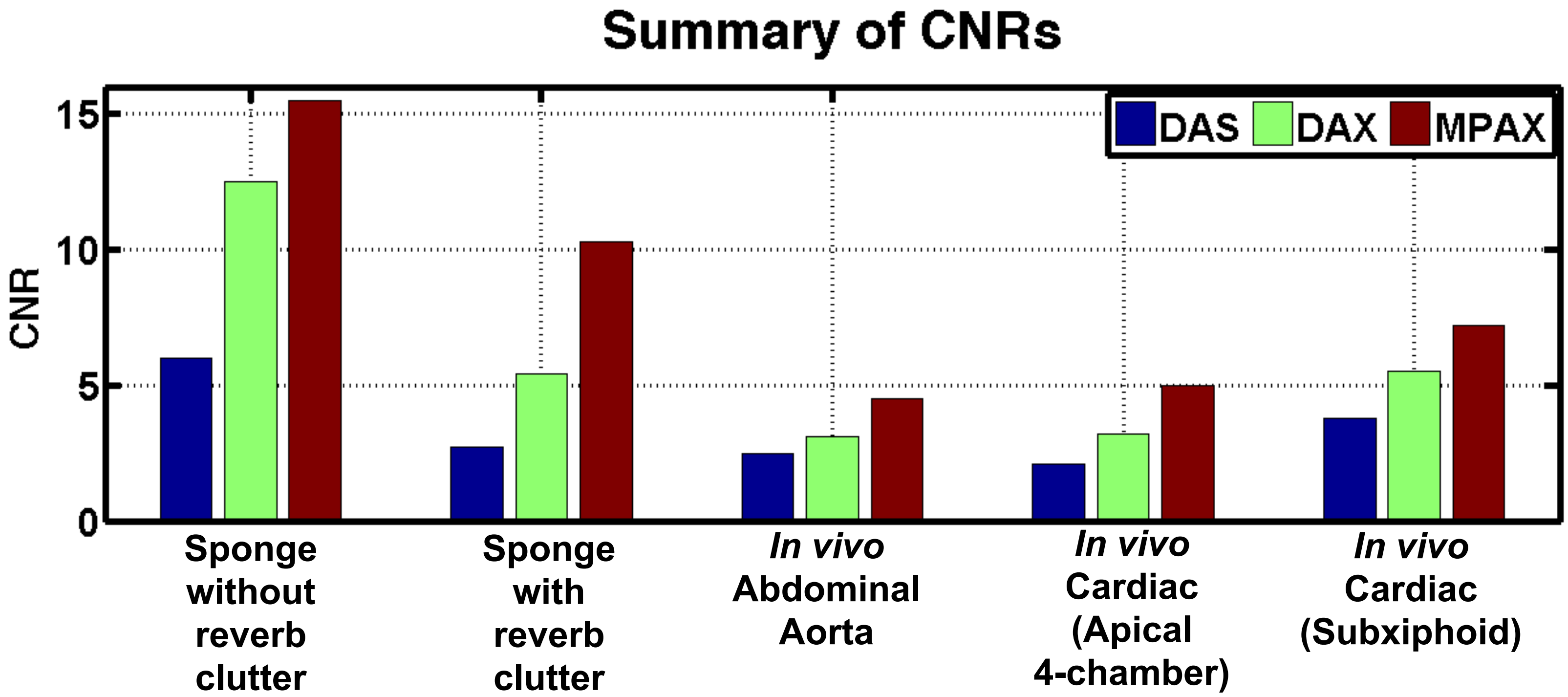
Subxiphoid View





		DAS	MPAX
CNR	**End-systole**	**3.9**	**7.2**
	End-diastole	**2.7**	**5.5**

MPAX: Conclusions



MPAX: Summary

- Highest CNR in all cases
- Robust with reverberation clutter
- Less prone to artifacts
- Better than or equivalent to other competing methods

Outline

I. **Image Quality Enhancement in Ultrasound Imaging**

   a. **Background**
   b. **Contrast Resolution: DAX & MPAX**
   c. **Signal-to-Noise Ratio: FXPF**
   d. **Spatial Resolution: MVBF & Deconvolution**

II. **Image Quality Enhancement in Ultrasound Tomography**

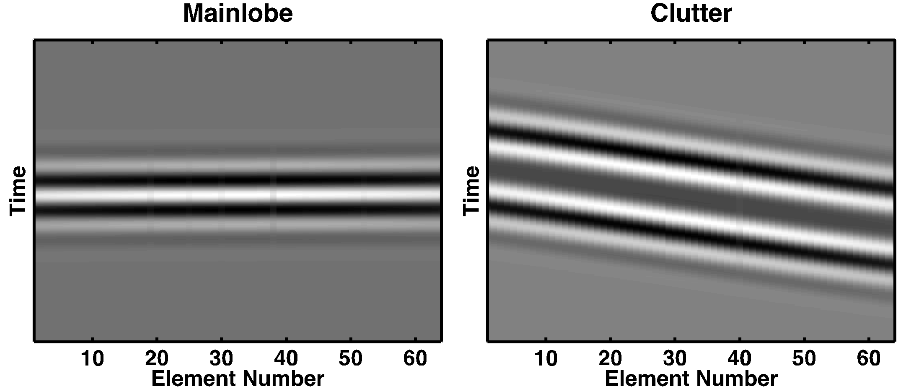
   a. **Background**
   b. **Sound Speed Reconstruction**

FX Prediction Filtering (FXPF)

- **Frequency-space (F-X) domain filtering for random noise suppression**
  - Linear/quasilinear events in time-space (T-X) domain → superposition of harmonics in F-X domain
- **Application to medical ultrasound imaging**
  - Coherent signals appear as linear events in the aperture domain
  - Incoherent signals (i.e. random noise and clutter) appear as random or pseudorandom

FXPF: Overview

Channel RF signals from a point target in T-X domain



Signals appear as linear events across the aperture

T-X Domain

- **Single linear event**

$$s_t(x+1) = s_{t-xg\psi}(1)$$

where g : Pitch of the transducer array

ψ : Slope of a linear event in the aperture domain

$s_t(x)$: RF signal from the x^{th} element at time *t*.

$$\Leftrightarrow$$

F-X Domain

$$S_f(x+1) = S_f(1)e^{-i2\pi f x g\psi}$$

For a specific frequency f_0, we obtain a linear recursion:

$$S_{f_0}(x+1) = a_{f_0}(1)S_{f_0}(x)$$

AR model of order 1

where $a_{f_0}(1) = e^{-i2\pi f_0 g\psi}$

FXPF: Overview

- Superposition of *p* linear events in the T-X domain can be represented by an AR model of order *p:*

$$S_{f_0}(x+1)=a_{f_0}(1)S_{f_0}(x)+a_{f_0}(2)S_{f_0}(x-1)+\ldots+a_{f_0}(p)S_{f_0}(x+1-p)$$

- Formulated as a convolutional form:

$$\mathbf{d}=\mathbf{f} * \mathbf{a}$$

where **a:** prediction filter of length *p*

f: vector containing $S_{f_0}(x)(x=1,2,3,\ldots,X)$

d: vector containing $S_{f_0}(x+1)(x=1,2,3,\ldots,X)$

- Reformulated as a matrix vector form:

$$\mathbf{d}=\mathbf{F}\mathbf{a}$$

This equation is based on a clean signal model.
In reality, the channel RF data are corrupted by random noise.

FXPF: Overview

- Solve for the prediction filter **a** from noise corrupted observation **d** based on the minimum prediction error energy assumption.

- Minimizing the following objective function:

$$J = \| \mathbf{Fa} - \mathbf{d} \|_2^2$$

We get:

$$\mathbf{F}^{\mathrm{T}}\mathbf{d} = \mathbf{F}^{\mathrm{T}}\mathbf{Fa}$$

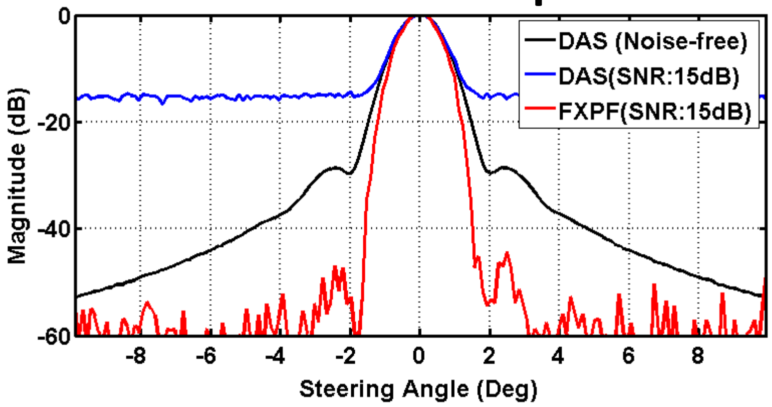
- The estimated clean data can be expressed as:

$$\hat{\mathbf{d}} = \mathbf{F}\hat{\mathbf{a}}$$

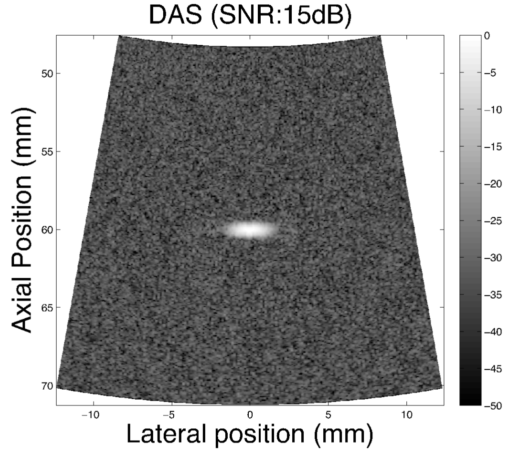
FXPF: Simulated Point Target Results

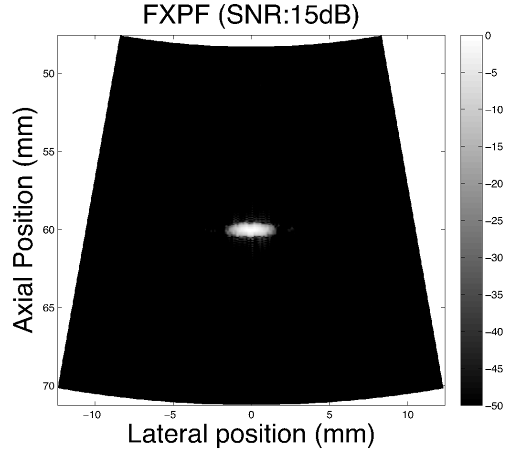
- **SNR: 15dB**

Lateral Beamplots

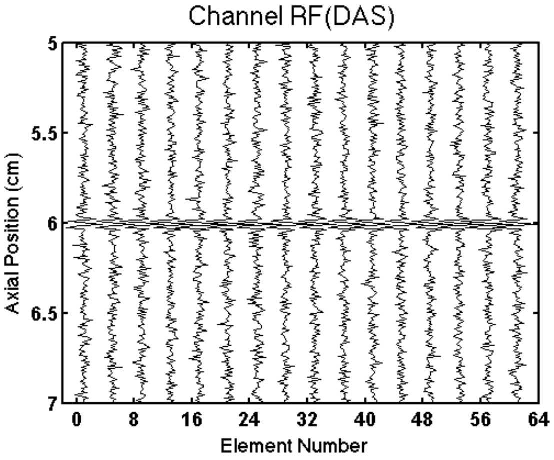


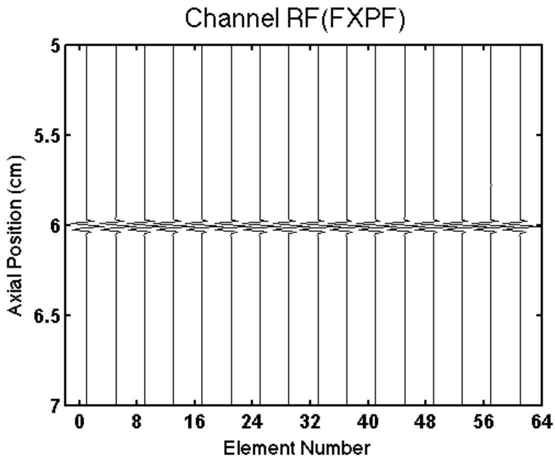
Beamformed Images





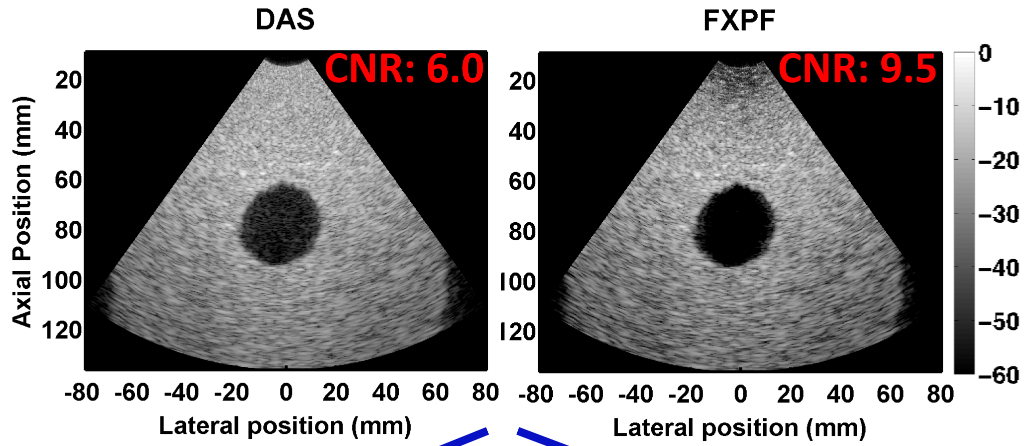
Channel RF data from x=0 mm





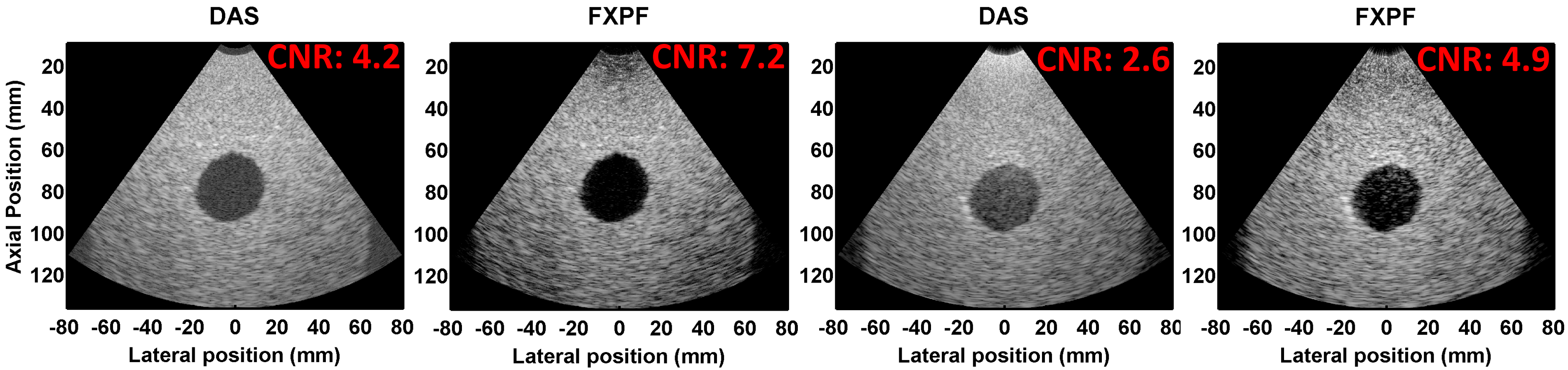
FXPF: Sponge Phantom Results

No Added Noise or Reverb Clutter



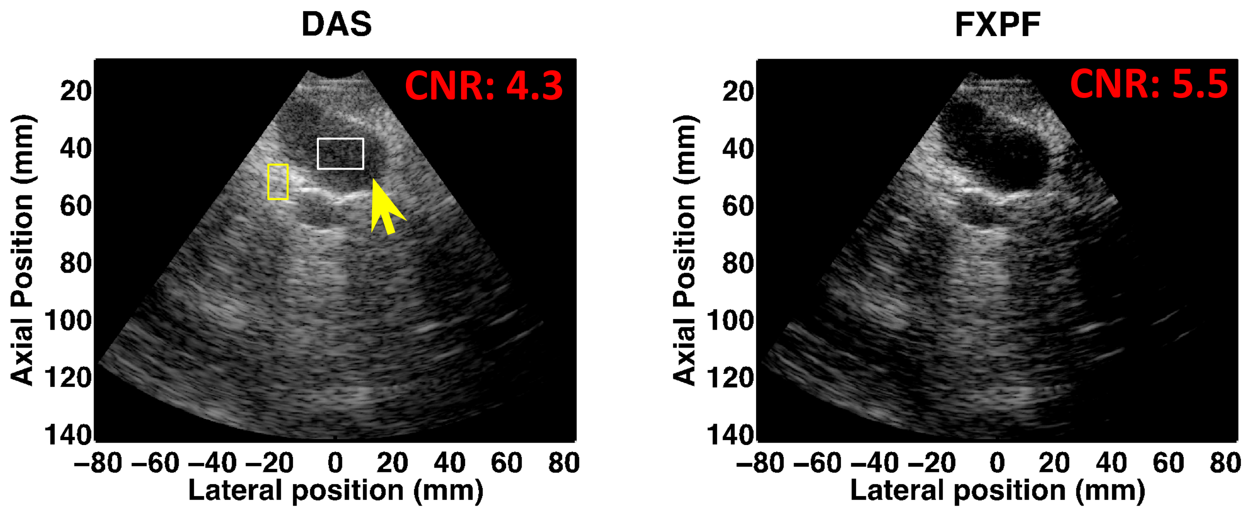
SNR: 15dB

Reverb Clutter

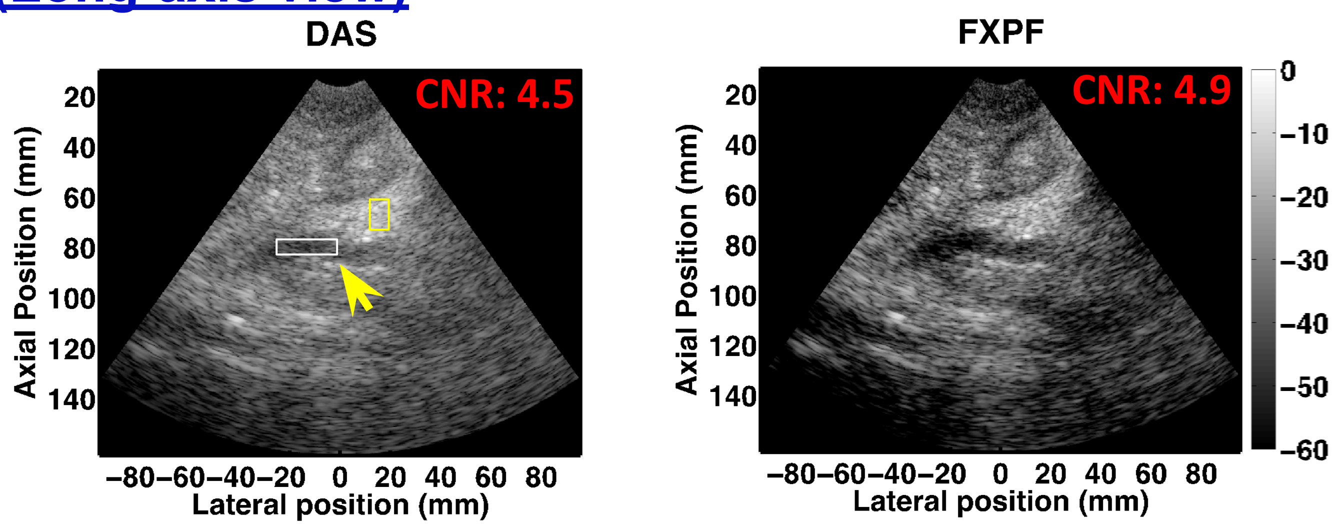


FXPF: *In vivo* Abdominal Results

- **Gall bladder (Long-axis view)**



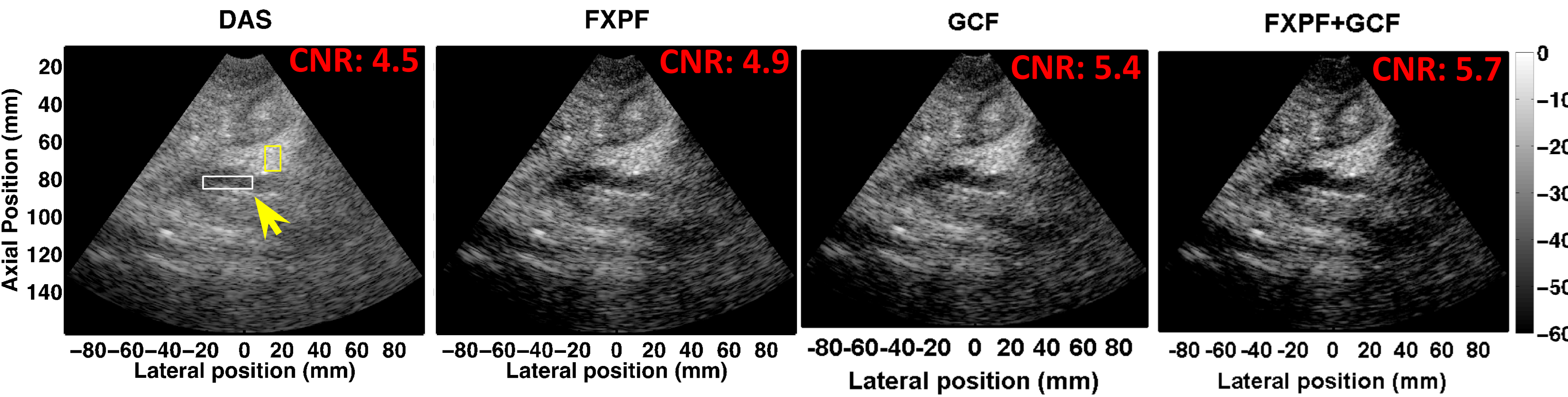
- **IVC (Long-axis view)**



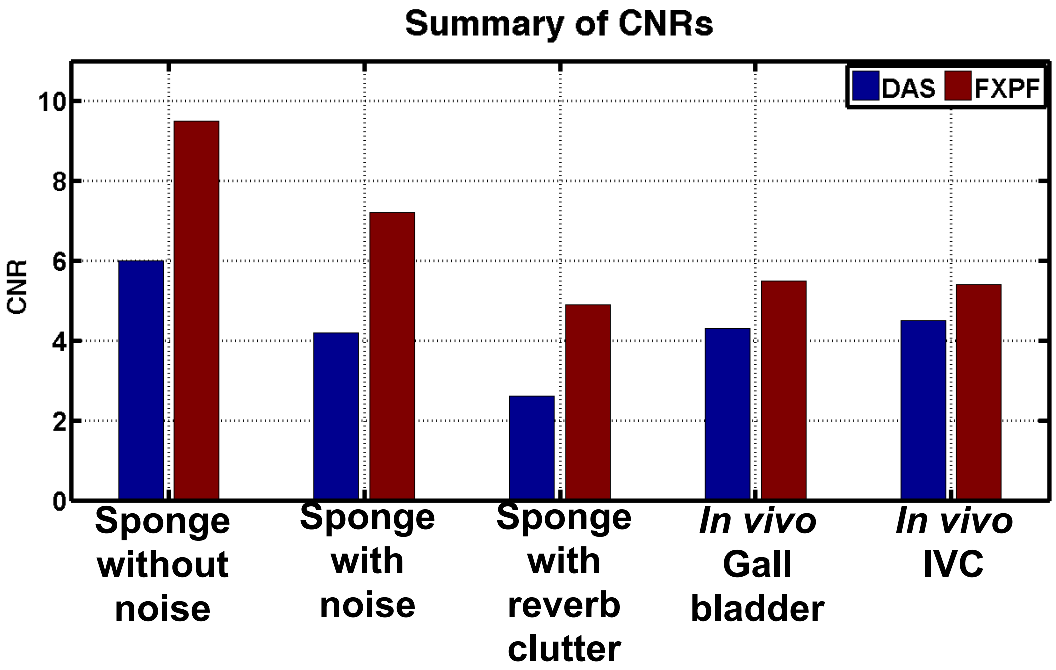
IVC: Inferior vena cava

Why FXPF?

- **Why another contrast enhancement technique?**
  - FXPF improves image contrast by enhancing channel RF SNR.
  - FXPF does not need to create a weighting matrix.
  - FXPF is based on a new mechanism. → Possibility for a hybrid approach



FXPF: Conclusions



FXPF Summary

- ✓ Suppresses any incoherent signals in the aperture domain
- ✓ Highly effective and robust
- ✓ Straightforward implementation
- ✓ Computationally efficient
- ✓ Does not create a weighting matrix as most other methods
- ✓ Can be applied iteratively

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   a. **Background**
   b. **Sound Speed Reconstruction**

Minimum Variance Beamforming (MVBF)

- Origins
  - J. Capon, "High-resolution frequency-wavenumber spectrum analysis," Proc. IEEE, pp. 1408-1418, 1969.
- Adaptive Beamformer
  - Data-dependent instead of predetermined aperture weights
- Main Benefits
  - Improved lateral resolution
  - Some sidelobe/clutter suppression

MVBF: Overview

- MVBF
  - Given DAS beamformer output $z[n]$:

$$z[n]=\mathbf{w}^H[n]\mathbf{X}[n] \quad \text{where} \quad \mathbf{w}[n]=\begin{bmatrix} w_1^*[n] \\ w_2^*[n] \\ \vdots \\ w_M^*[n] \end{bmatrix} \quad \text{and} \quad \mathbf{X}[n]=\begin{bmatrix} x_1[n-\Delta_1[n]] \\ x_2[n-\Delta_2[n]] \\ \vdots \\ x_M[n-\Delta_M[n]] \end{bmatrix}$$

  - The variance of $z[n]$ can be written as:

$$E\left[\left|z[n]\right|^2\right]=\mathbf{w}^H[n]\mathbf{R}[n]\mathbf{w}[n]$$

$$\mathbf{R}[n]=E\left[\mathbf{X}[n]\mathbf{X}^H[n]\right]\text{: Spatial covariance matrix}$$

  - Minimize the variance of $z[n]$ while forcing unit gain at the focal point:

$$\min_{\mathbf{w}[n]} \mathbf{w}^H[n]\mathbf{R}[n]\mathbf{w}[n]$$

$$\text{subject to } \mathbf{w}^H[n]\mathbf{a}=1 \qquad \mathbf{a}\text{ : Steering vector}$$

MVBF: Overview

- MVBF
  - The optimization problem has an analytical solution:

$$\mathbf{w}[n] = \frac{\mathbf{R}^{-1}[n]\mathbf{a}}{\mathbf{a}^{H}\mathbf{R}^{-1}[n]\mathbf{a}}$$

  - $\mathbf{R}[n]$ must be estimated by averaging in spatial and temporal domain:

$$\hat{\mathbf{R}}[n] = \frac{1}{(2K+1)(M-L+1)} \sum_{k=-K}^{K} \sum_{l=1}^{M-L+1} \bar{\mathbf{X}}_l[n-k]\bar{\mathbf{X}}_l^{H}[n-k] \quad \text{where } \bar{\mathbf{X}}_l[n] = \begin{bmatrix} x_l[n] \\ x_{l+1}[n] \\ \vdots \\ x_{l+L-1}[n] \end{bmatrix}$$

Spatial averaging over M-L+1 subarrays

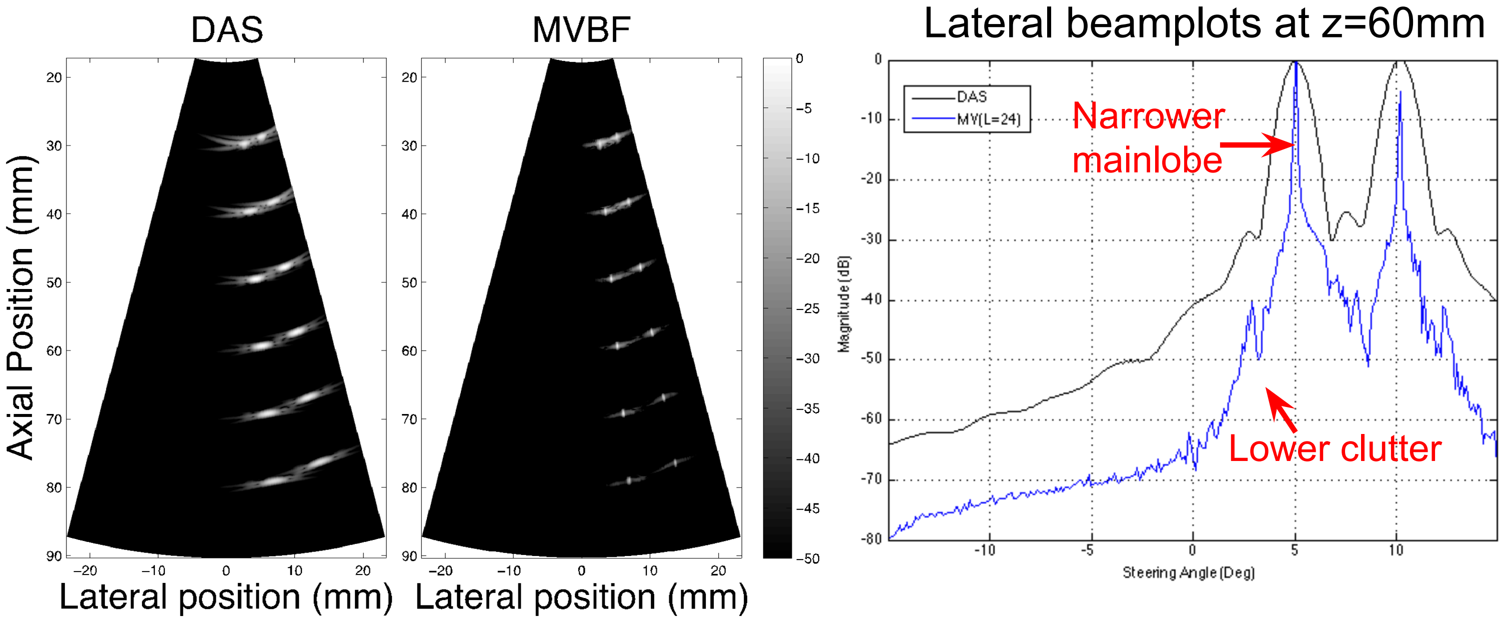
Temporal averaging over 2K+1 samples

  - The final MV amplitude estimate is:

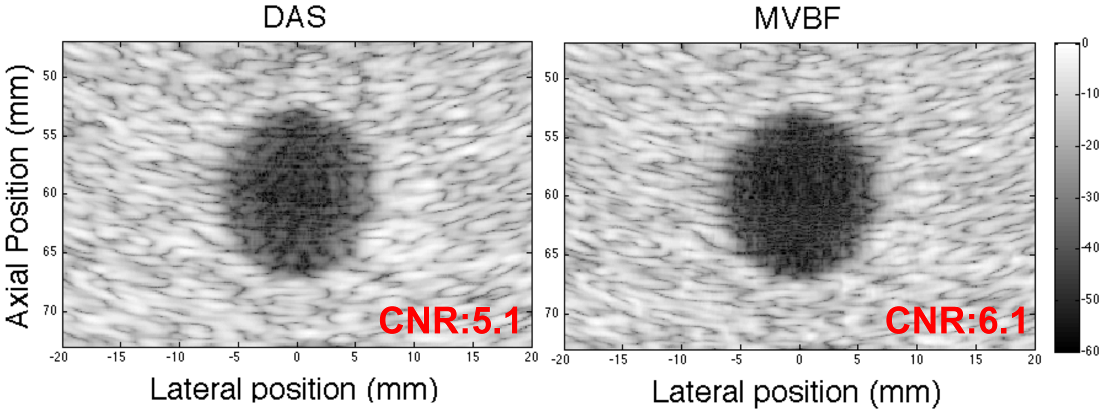
$$z_{MV}[n] = \frac{1}{M-L+1} \sum_{l=1}^{M-L+1} \mathbf{w}^{H}[n]\bar{\mathbf{X}}_l[n]$$

MVBF: Simulation Examples

Simulated Point Targets



Simulated Anechoic Cysts



Spiking Deconvolution

- Goal
  - Aims to “sharpen” the channel RF signals by removing the effect of ultrasound pulse using an inverse filter estimated from the data itself.

- Main Benefits
  - Improved axial resolution
  - Broader frequency spectrum
  - Slightly enhanced contrast

Spiking Deconvolution: Overview

- Define error between desired & actual outputs

$$E = \sum_t (d_t - y_t)^2 = \sum_t (d_t - f_t * x_t)^2$$

d_t :Desired output
y_t :Actual output

- Minimizing the error *E*, we get:

$$\begin{pmatrix} r_0 & r_1 & r_2 & \cdots & r_{n-1} \\ r_1 & r_0 & r_1 & \cdots & r_{n-2} \\ \vdots & \vdots & \vdots & \ddots & \vdots \\ r_{n-1} & r_{n-2} & r_{n-3} & \cdots & r_0 \end{pmatrix} \begin{pmatrix} f_0 \\ f_1 \\ \vdots \\ f_{n-1} \end{pmatrix} = \begin{pmatrix} g_0 \\ g_1 \\ \vdots \\ g_{n-1} \end{pmatrix}$$

r_i : Autocorrelation lags of the input

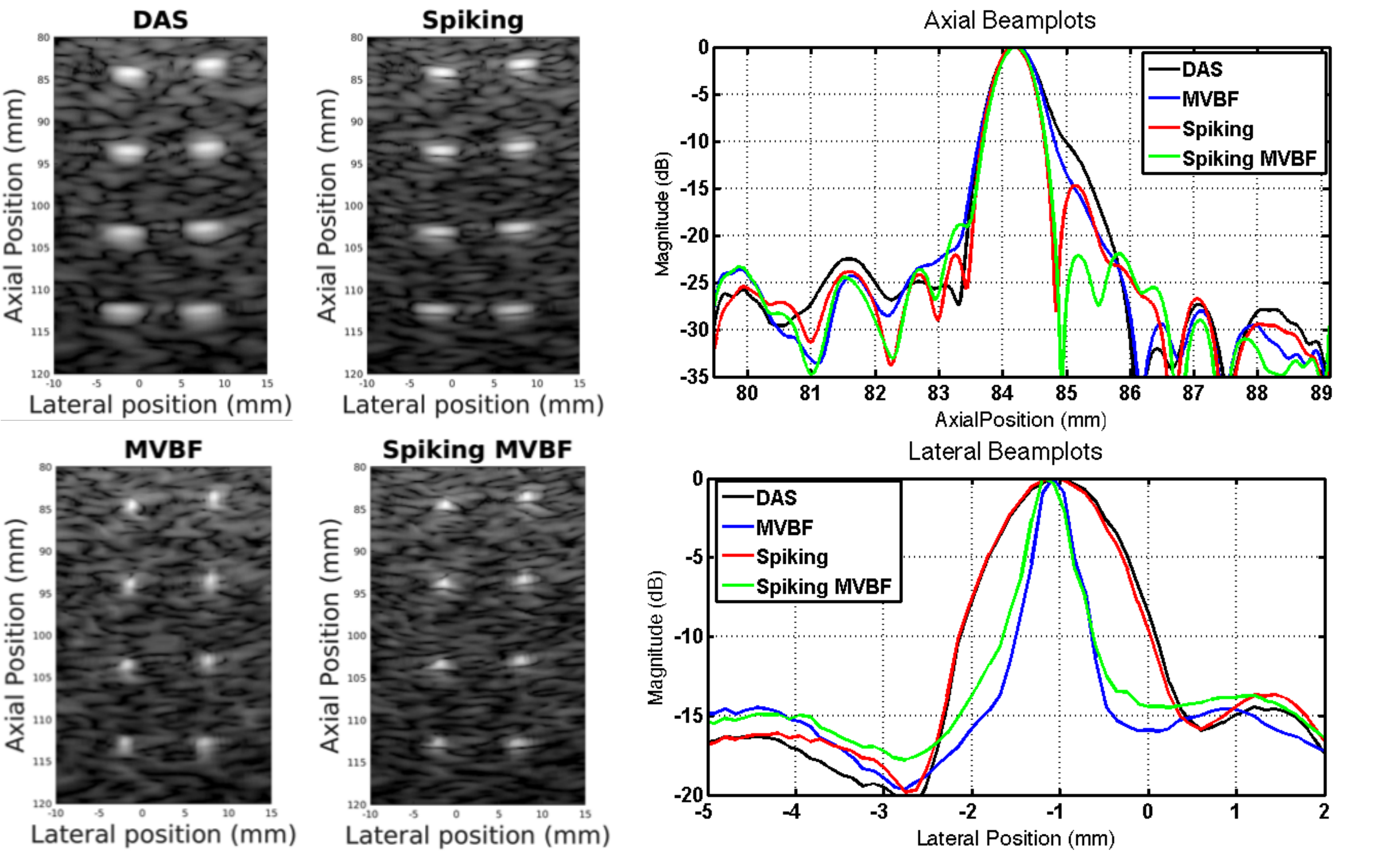
g_i : Cross-correlation lags between desired output & input

- Given the input series $x_t : (x_0, x_1, x_2, \ldots)$, find the inverse filter such that the desired output is a zero-lag spike $d_t : (1, 0, 0, \ldots)$:

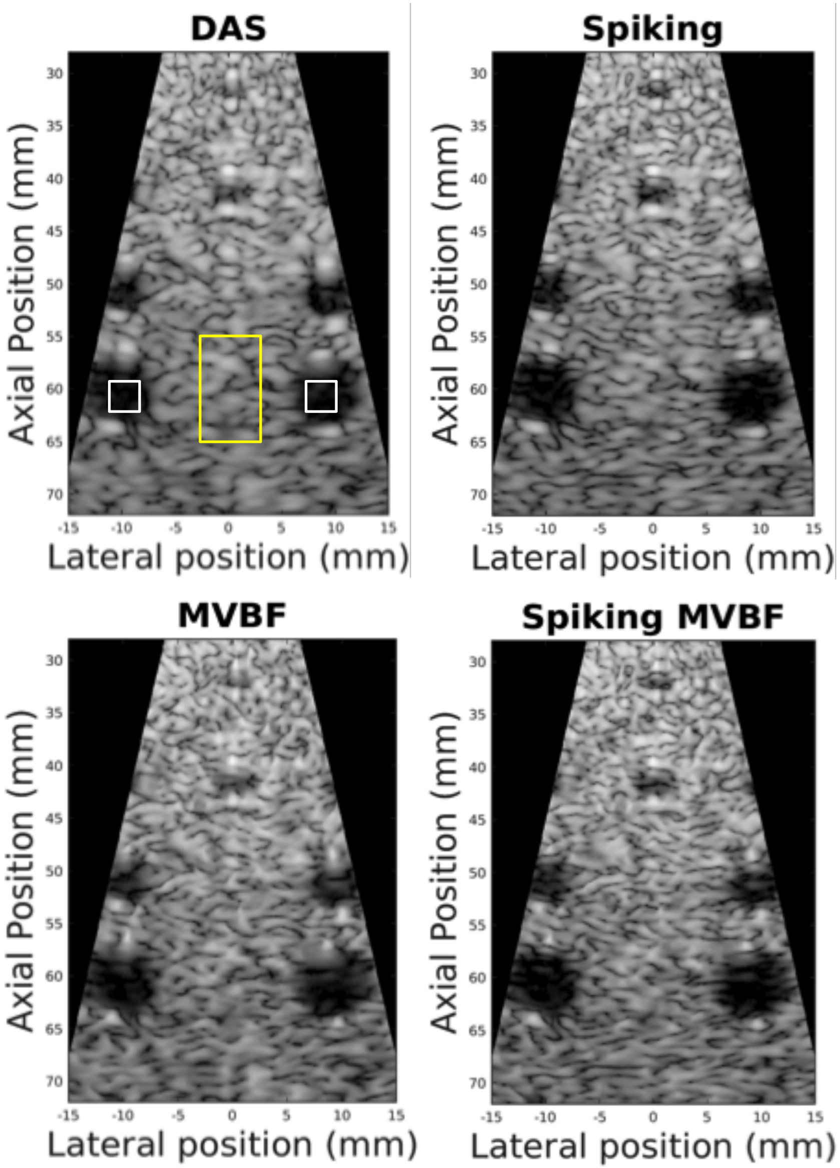
$$\begin{pmatrix} r_0 & r_1 & r_2 & \cdots & r_{n-1} \\ r_1 & r_0 & r_1 & \cdots & r_{n-2} \\ \vdots & \vdots & \vdots & \ddots & \vdots \\ r_{n-1} & r_{n-2} & r_{n-3} & \cdots & r_0 \end{pmatrix} \begin{pmatrix} f_0 \\ f_1 \\ \vdots \\ f_{n-1} \end{pmatrix} = \begin{pmatrix} x_0 \\ 0 \\ \vdots \\ 0 \end{pmatrix}$$

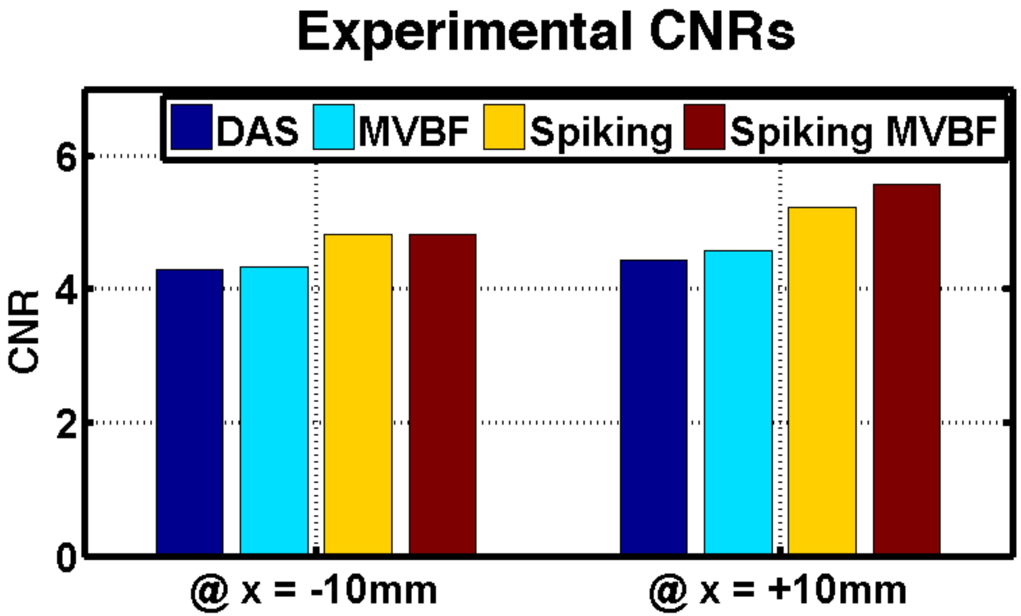
- Filter the data using the inverse filter *f*

Experimental Results: Beamplots



Experimental Results: Contrast





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   a. **Background**
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Breast Cancer Background

Breast Cancer

- One of the most common types of cancers in women
- Early detection → higher survival rate
- Gold standard for breast cancer screening: mammography
- Many recent studies are skeptical about the benefit of mammography

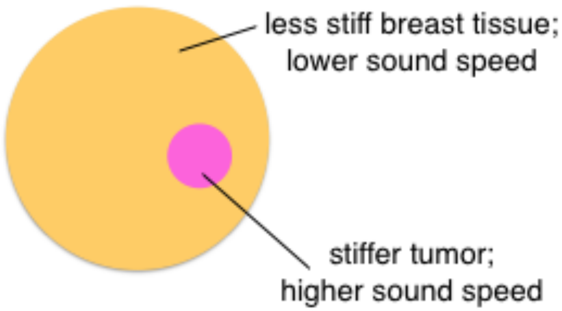
Risks of mammography

- Ionizing radiation
- Difficulty discerning benign vs. malignant
- Difficulty imaging dense breasts
- False-negative results miss cancer
- Discomfort, pain

https://thenypost.files.wordpress.com/

Breast Cancer Background

Characterizing tumors based on sound speed



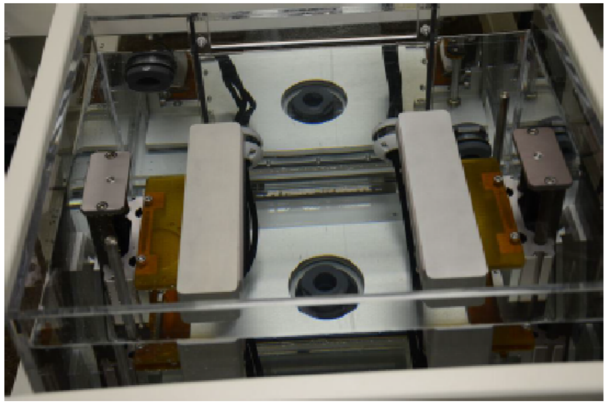
Breast tissue type	Sound speed (m/s)
Fatty tissue	1422 ± 9
Glandular (dense) tissue	1487 ± 21
Benign lesion	1513 ± 27
Malignant lesion	1548 ± 17

Li et al, Ultrs Med Bio, 2009

Breast Ultrasound Tomography

Ultrasound Tomography System

(a) Breast ultrasound tomography prototype.

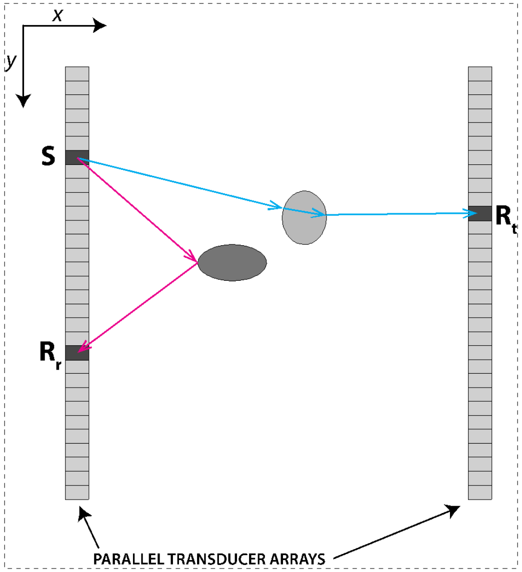


(b) Two parallel ultrasound transducer arrays.

Two parallel transducer arrays:

- Fits variable breast size
- Can image underarm region

Ultrasound Bent-ray Tomography (USRT):



- USRT uses first arrival times of the transmission (and possibly reflection) signals for tomographic reconstruction .

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USRT: The Forward Problem

Eikonal equation:

$$\left(\frac{\partial t}{\partial x}\right)^2 + \left(\frac{\partial t}{\partial y}\right)^2 = \left(\frac{1}{v}\right)^2 = (s_x^2 + s_y^2)$$

v : sound speed
t : travel time
(s_x, s_y) : slowness vector

Time-slowness relationship:

$$t_i = \sum_{j=1}^{N} l_{ij}/v_j = \sum_{j=1}^{N} l_{ij} s_j$$

v_j : sound speed in j^th cell

t_i : time

l_{ij} : length of i^th ray path in j^th cell

s_j : slowness in j^th cell

N : total # cells in model

Matrix form:

$$\mathbf{T} = H\mathbf{s}$$

$\mathbf{T}$: travel time vector

H : matrix of ray path segments l_{ij}

$\mathbf{s}$: slowness vector

USRT: Regularized Inversion

Minimization problem:

$$E(\mathbf{s}) = \min_{\mathbf{s}}\{\|H\mathbf{s} - \mathbf{D}\|_2^2\}$$

D : data vector of observed times

Hs : forward modeling result

Minimization problem with Tikhonov regularization:

$$E(\mathbf{s}) = \min_{\mathbf{s}}\{\|H\mathbf{s} - \mathbf{D}\|_2^2 + \lambda\|\mathbf{L}\mathbf{s}\|_2^2\}$$

λ : regularization parameter

L : regularization operator

Minimization problem with TV regularization:

$$E(\mathbf{s}) = \min_{\mathbf{s}}\{\|H\mathbf{s} - \mathbf{D}\|_2^2 + \lambda_{TV}\|\mathbf{s}\|_{TV}\}$$

where $$\|\mathbf{s}\|_{TV} = \sum_{i,j=1}^{n}\sqrt{|(\nabla_x \mathbf{s})_{i,j}|^2 + |(\nabla_y \mathbf{s})_{i,j}|^2}$$

$\boldsymbol{\lambda}_{TV}$: TV regularization parameter

$\nabla_x \mathbf{s}_{i,j}$: spatial derivative along x

$\nabla_y \mathbf{s}_{i,j}$: spatial derivative along y

Minimization problem with MTV regularization:

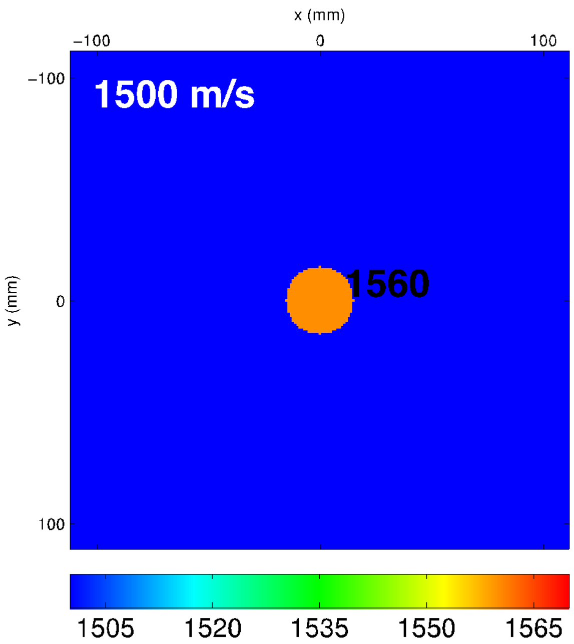
$$E(\mathbf{s}) = \min_{\mathbf{s},\mathbf{u}}\{\|H\mathbf{s} - \mathbf{D}\|_2^2 + \lambda_1\|\mathbf{s} - \mathbf{u}\|_2^2 + \lambda_2\|\mathbf{u}\|_{TV}\}$$

$\boldsymbol{\lambda}_1, \boldsymbol{\lambda}_2$: regularization parameters

u : second minimization parameter

Numerical Phantom Results 1

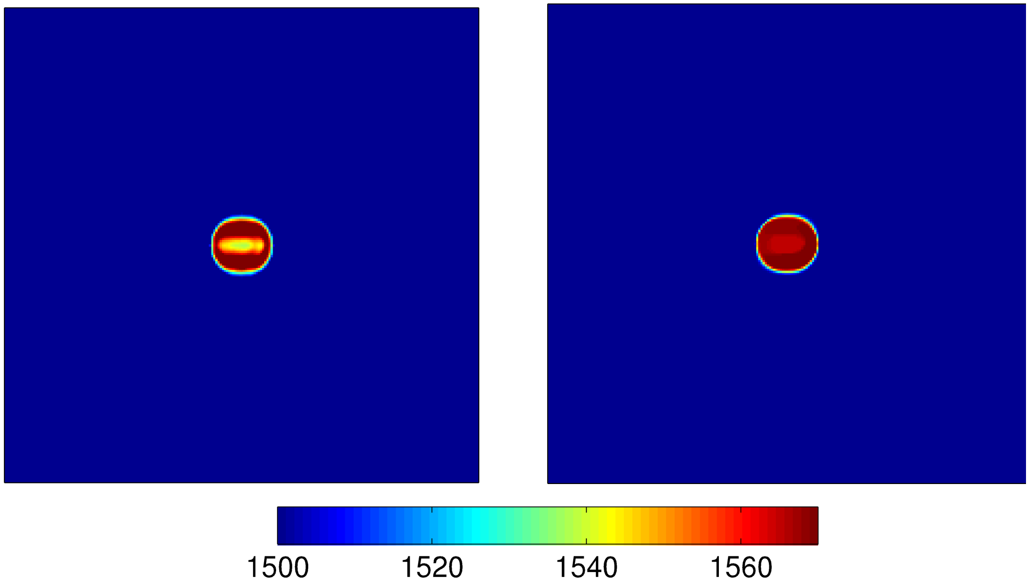
True Phantom



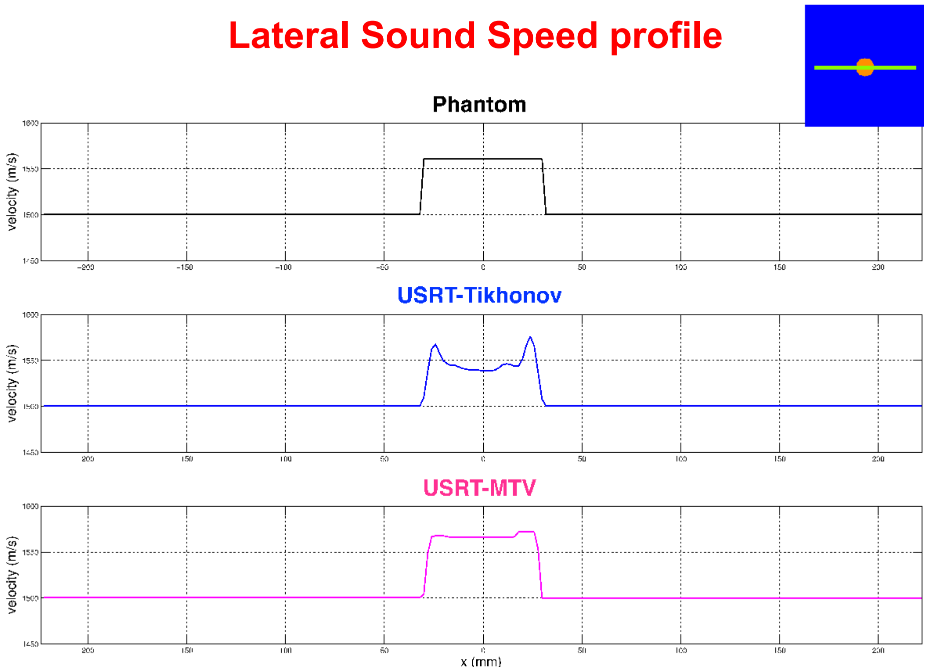
USRT Reconstruction

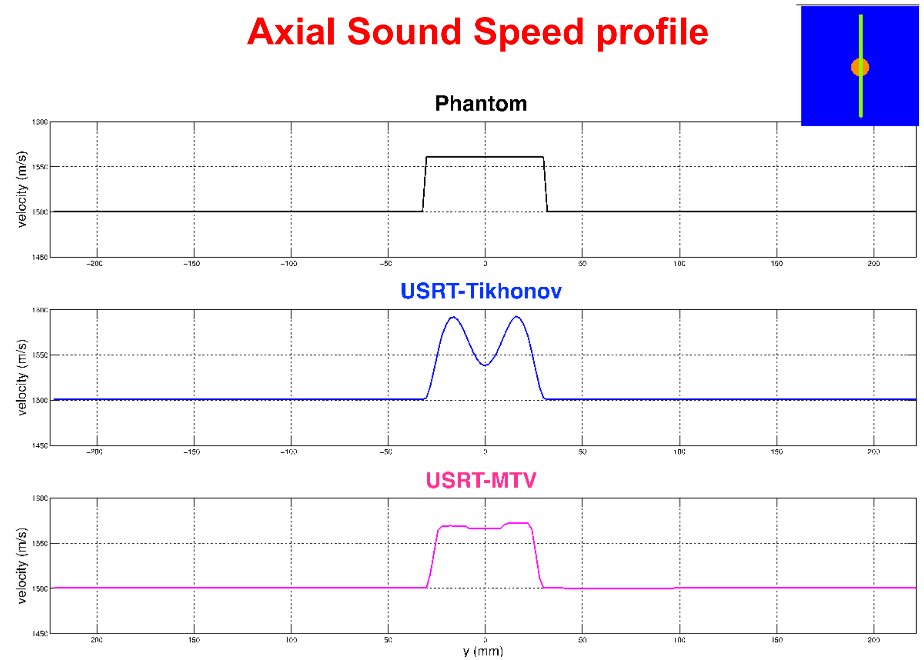
Tikhonov

MTV



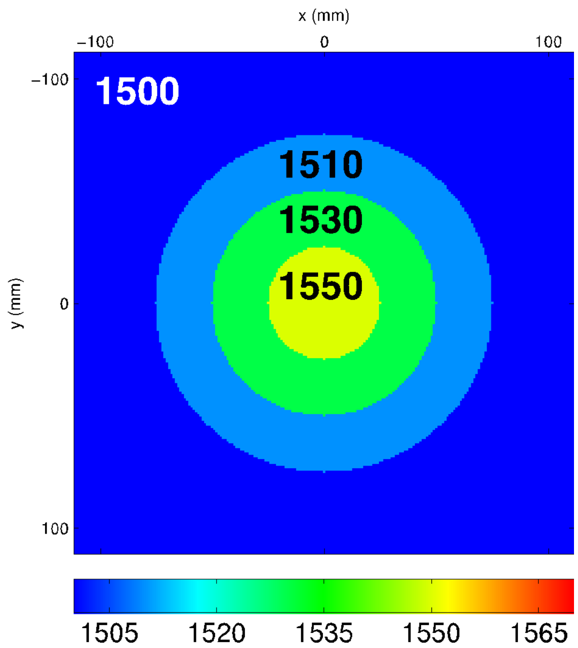
Numerical Phantom Results 1





Numerical Phantom Results 2

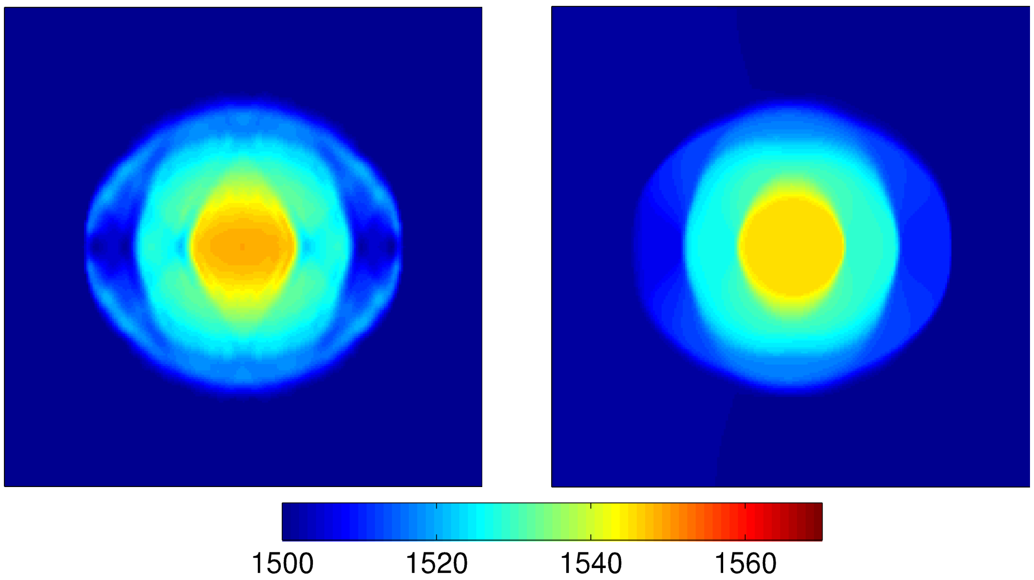
True Phantom



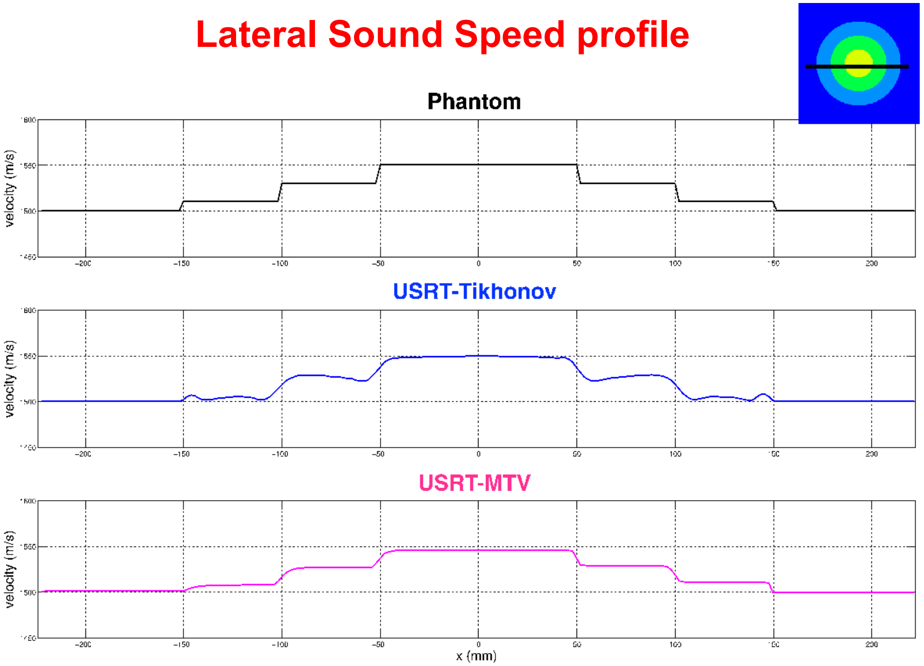
USRT Reconstruction

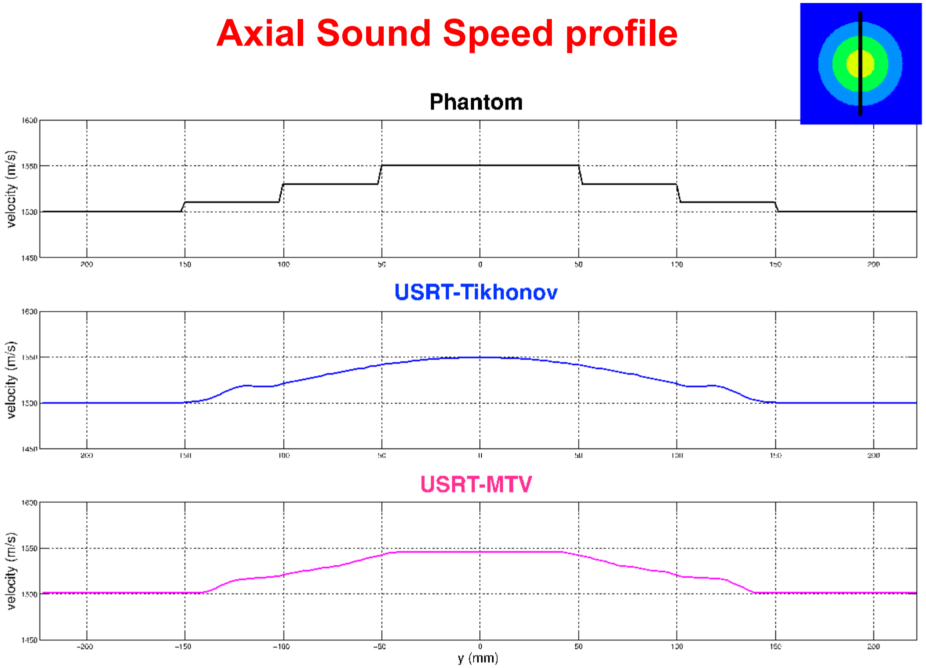
Tikhonov

MTV



Numerical Phantom Results 2





USRT: Conclusions

USRT Summary

- ✓ Developed USRT with MTV-regularization
- ✓ Demonstrated improvement in sound speed reconstruction with simulated data
- ✓ Currently, the new algorithm is being is being validated with phantom and *in vivo* patient data

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Acknowledgments

- Jesse Yen, PhD (USC)
- Lianjie Huang, PhD (LANL)
- Kenneth Hanson, PhD (LANL)
- Youzuo Lin, PhD (LANL)
- Ting Chen, PhD (LANL)
- Miranda Intrator, MS (LANL)
- Man Nguyen, PhD (Philips)