# *Synthetic Aperture Ultrasound Imaging*
## *Part 1: B-mode imaging*

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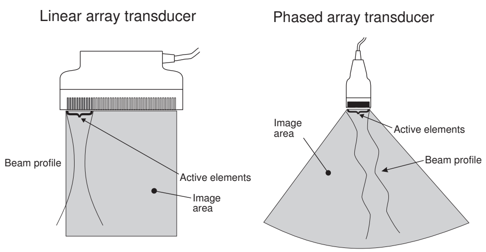
## Principles of Synthetic Aperture Imaging - Outline

- Problems in modern ultrasound scanners.
  Need for change of the acquisition
- What is synthetic aperture imaging?
- Acquisition of synthetic aperture images
- Equations for resolution, signal-to-noise ratio.
  Fundamental limitations
- Changes necessary to use it in ultrasound



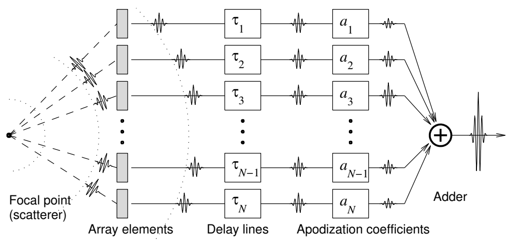
## Conventional Ultrasound Scanners

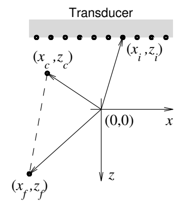
- Multielement transducer arrays are used:
  - Linear arrays - element width $\approx \lambda$
  - Phased arrays - element width $\approx \lambda/2$
- A beam is formed in transmit by applying different delays on the transducer elements
- As ultrasound pulse propagates along the beam direction echoes are scattered back to the transducer.
- The transducer elements convert the received echo back to electrical signal.
- The electrical signals are appropriately delayed. The delays change as a function of time (dynamic delays).
- The delayed signals are summed forming an *A-line*.
- The same process is repeated for another image direction.






## Beamforming in Modern Scanners




$$s(t) = \sum_{1}^{N_{xdc}} a_i r_i(t - \tau_i)$$

$$\tau_i = \frac{|\vec{x}_c - \vec{x}_f| - |\vec{x}_i - \vec{x}_f|}{c}$$

- $a_i$ Weighting coefficient (apodization)
- $r_i(t)$ Received signal
- $\vec{x} = [x, y, z]^T$ Spatial position
- $\vec{x}_i, \vec{x}_c, \vec{x}_f$ Position of transducer element, beam origin, and current focal point, respectively
- $c$ Speed of sound

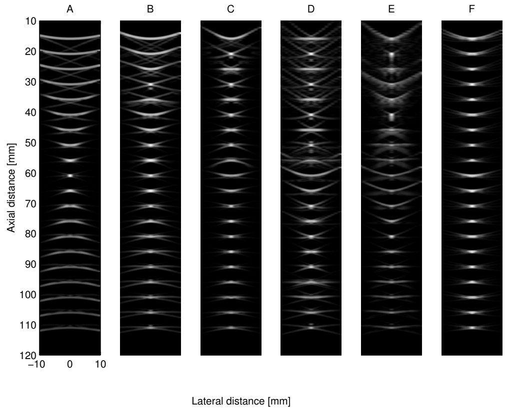
# System Characterization

A system can be characterized by the *point-spread-function* (PSF). The point spread function is:

$$\mathbf{p}(\vec{x},t) = v_{pe}(t) \underset{t}{*} h_{pe}(\vec{x},t).$$

Examples of PSF without apodization:

- A - one focal zone
- B - 6 receive focal zones
- C - 6 zones receive xmt and rcv
- D - 128 elem, 4 zones xmt, 7 zones rcv
- E - 128 elem, 4 zones xmt, dynamic rcv
- F - 128 elem, xmt $F_\# = 4$, rcv $F_\# = 2$



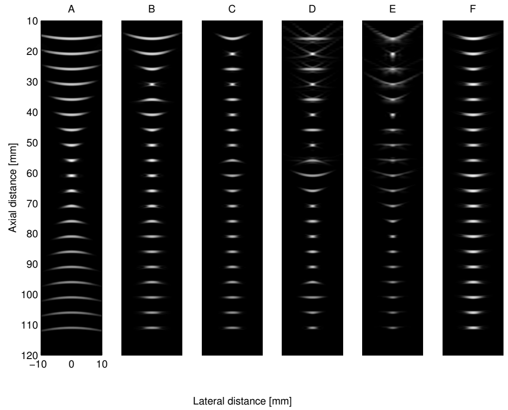



# Sidelobe Reduction

The sidelobes can be improved by applying apodization.

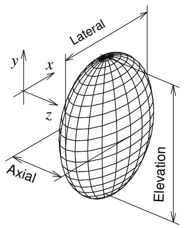
Examples of PSF with apodization:

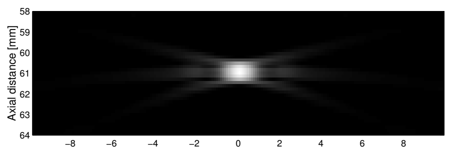
- A - one focal zone
- B - 6 receive focal zones
- C - 6 zones receive xmt and rcv
- D - 128 elem, 4 zones xmt, 7 zones rcv
- E - 128 elem, 4 zones xmt, dynamic rcv
- F - 128 elem, xmt $F_\# = 4$, rcv $F_\# = 2$

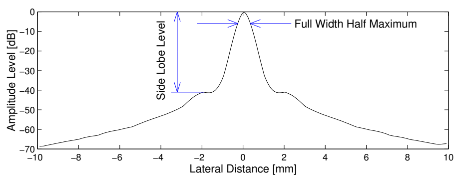





# PSF Characteristics




- The PSF is three dimensional
- The B-mode images are only 2-D
- Displayed on a logarithmic scale
- Maximum taken along $z$
- Parameters used: FWHM, side- and grating-lobe level

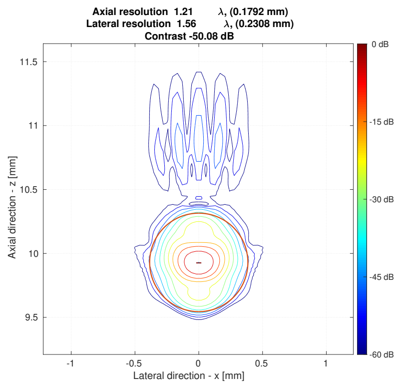


# PSF Contrast

Contrast ratio between energy outside main lobe normalized by energy of PSF

$$CR(r) = 20\log_{10}\sqrt{\frac{E_{out}(r)}{E_{total}}},$$

- $E_{total}$ total energy in point spread function
- $E_{out}(r)$ energy beyond a radius of $r$
- $E_{out}(r)$ calculated for a radius equal to $2.5\lambda$




PSF of linear array. 6 dB between contours.

Red circle indicates boundary between main and side-lobe

## Field for arrays - How do we determine the arrays geometry?

Linear medium, individual spatial impulse responses are summed:

$$h_a(\vec{r}_p,t) = \sum_{i=0}^{N-1} h_e(\vec{r}_i,\vec{r}_p,t),$$

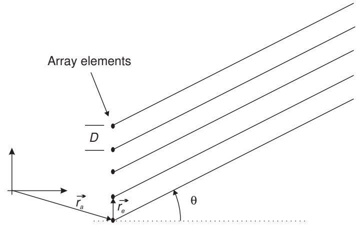
Assume elements are very small and field point is far away from the array:

$$h_a(\vec{r}_p,t) = \frac{k}{R_p}\sum_{i=0}^{N-1} \delta(t-\frac{|\vec{r}_i-\vec{r}_p|}{c})$$

Note, spherical wave.
$R_p$ - Distance to transducer
$k$ - Constant of proportionality



## Array geometry




*Geometry of linear array*

If spacing between elements is $D$, then

$$h_a(\vec{r}_p,t) = \frac{k}{R_p}\sum_{i=0}^{N-1} \delta\left(t-\frac{|\vec{r}_a+iD\vec{r}_e-\vec{r}_p|}{c}\right)$$

Difference in arrival time between elements far from the transducer is

$$\Delta t = \frac{D\sin\Theta}{c}.$$

Combined spatial impulse response is, thus, a series of Dirac pulses separated by $\Delta t$.

$$h_a(\vec{r}_p,t) \approx \frac{k}{R_p}\sum_{i=0}^{N-1} \delta\left(t-\frac{R_p}{c}-i\Delta t\right)$$



## Beam pattern

Beam pattern as a function of angle is found by Fourier transforming $h_a$

$$\begin{aligned} H_a(f) &= \frac{k}{R_p}\sum_{i=0}^{N-1} \exp\left(-j2\pi f\left(\frac{R_p}{c}+i\frac{D\sin\Theta}{c}\right)\right) \\ &= \exp(-j2\pi\frac{R_p}{c})\frac{k}{R_p}\sum_{i=0}^{N-1} \exp\left(-j2\pi f\frac{D\sin\Theta}{c}\right)^i \\ &= \frac{\sin(\pi f\frac{D\sin\Theta}{c}N)}{\sin(\pi f\frac{D\sin\Theta}{c})}\exp\left(-j\pi f(N-1)\frac{D\sin\Theta}{c}\right)\frac{k}{R_p}\exp(-j2\pi\frac{R_p}{c}). \end{aligned}$$

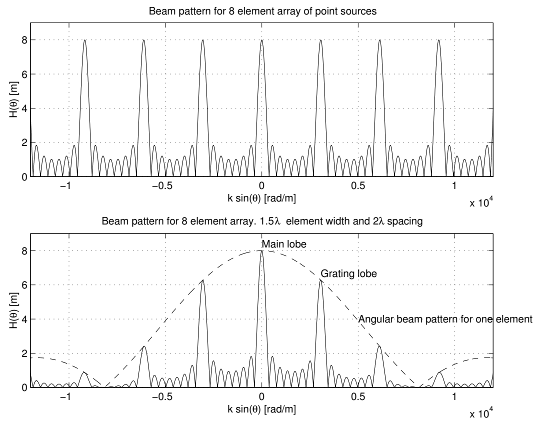
Amplitude of the beam profile:

$$|H_a(f)| = \left|\frac{k}{R_p}\frac{\sin(\pi N\frac{D}{\lambda}\sin\Theta)}{\sin(\pi\frac{D}{\lambda}\sin\Theta)}\right|.$$

Note correspondence to Fourier transform of digital square wave.



## Continuous wave field of point sources array




Grating lobes for array consisting of 8 point elements (top) and of 8 elements with a size of 1.5λ (bottom). The pitch (distance between elements) is 2λ.

## Interpretation and consequences

Beam profile:

$$|H_a(f)| = \left| \frac{k}{R_p} \frac{\sin(\pi N \frac{D}{\lambda} \sin\Theta)}{\sin(\pi \frac{D}{\lambda} \sin\Theta)} \right|$$

$D$ - Pitch of transducer.
$N$ - Number of elements.
$ND$ - Width of array.

Main lobe at $\Theta = 0$ or $n = 0$. Width from zeros at:

$$N \frac{D \sin\Theta}{\lambda} = 1 \Rightarrow \Theta_w = 2 \arcsin \frac{\lambda}{ND}$$

Other peaks should be avoided.
Poles in transfer function:

$$\frac{D \sin\Theta}{\lambda} = n$$

$n$ - Integer $\neq 0$.
Corresponds to peaks in the beam pattern.
Demand for no grating lobe:

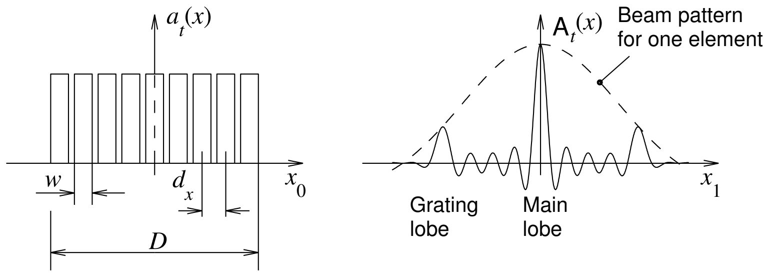
$$\frac{D \sin\Theta}{\lambda} < 1 \Rightarrow D < \frac{\lambda}{\sin\Theta}$$

For linear array: $D < \lambda$.
For phased array: $D < \lambda/2$ for safety margin for beam steering.



## Fourier Relations

Apodization $a_t(x)$ related to radiation pattern $A_t(x)$ through a Fourier relation.




$$a_t(x_0) = \Pi(x_0/D) \left[ \Pi(x_0/w) \underset{x}{*} \mathrm{III}(x_0/d_x) \right]$$

$$A_t(x_1) = \mathrm{sinc}\left(\frac{x_1}{\lambda z_1} w\right) \cdot \left[ \mathrm{sinc}\left(\frac{x_1}{\lambda z_1} D\right) \underset{x}{*} \mathrm{III}\left(\frac{x_1}{\lambda z_1} d_x\right) \right]$$

Element width Array size Pitch of array



## Improvement Sought for Conventional Imaging

### Acquisition speed

- One line scanned per transmission
- One transmission lasts $\sim 200$ $\mu$s for a typical scan depth of $z_{max} = 15$ cm and speed of sound $c = 1540$ m/s.
- Typically 1.5 to 2 lines per degree or per element, giving up to 120 – 200 lines per image.
- Frame rates 20 – 40 frames per second (fps).

### Image quality

- The focus in transmit is fixed $\rightarrow$ poor resolution before and after the focus.
- Using several transmissions and "stitching" the image improves the quality. The frame rate falls down to 5 – 10 fps.

### Flow estimation

- 8 – 16 emissions are used to estimate the flow in one direction
- The frame rate is decreased. The estimates are inaccurate.



## Synthetic Aperture Focusing (SAF) - History

- Appeared in 1950s and first applied to radar systems
- Purpose: synthesize a large array to increase resolution
- Applied to ultrasound since the end of 1970s
- Used mostly in non-destructive testing
- A lot of research in the 1990s for medical imaging.
- Applied in intra-vascular ultrasound. Improvement of resolution
- Since 2001 CFU has in-vivo flow estimation

## SAF - classification

In Synthetic Aperture Radar (SAR) the classification is based on the available number of antennas (physical apertures) :

- Mono-static - one antenna transmits and receives
- Bistatic - the transmitting and receiving antennas are different

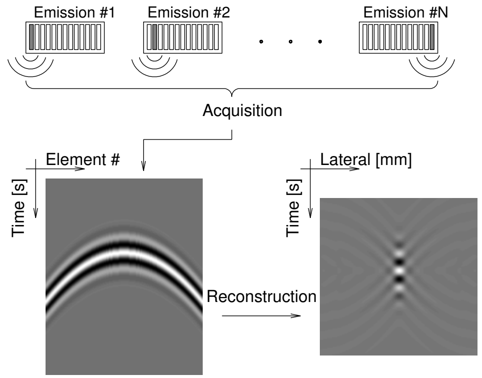
Synthetic Aperture Ultrasound (SAU) classification based on which aperture is being synthesized:

- Synthetic Transmit Aperture (STA) imaging
- Synthetic Receive Aperture (SRA) imaging
- Synthetic Transmit and Receive Aperture (STRA) imaging



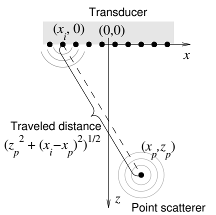
## Monostatic SAF - Acquisition

- One element transmits
- The same element receives
- The *cylindrical* wave covers the slice under investigation
- The data from all acquisitions is stacked together
- The reconstruction is performed either in frequency or in time domain.






## The received signal

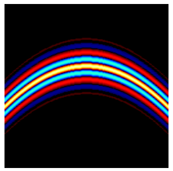



Propagation time:

$$t_p = \frac{2}{c}|\vec{r}_p - \vec{r}_i|$$

$$t_p = \frac{2}{c}\sqrt{(x_i - x_p)^2 + (z_i - z_p)^2}$$

$$t_p = \frac{2}{c}\sqrt{(x_i - x_p)^2 + z_p^2}$$



The received signal:

$$r_i(t) = \sigma_p g\left(t - \frac{2\sqrt{(x_i - x_p)^2 + z_p^2}}{c}\right) \quad \text{One scatterer}$$

$$r_i(t) = \sum_p \sigma_p g\left(t - \frac{2\sqrt{(x_i - x_p)^2 + z_p^2}}{c}\right) \quad \text{Multiple scatterers}$$

$g(t)$ - transmitted pulse



## Types of Reconstruction

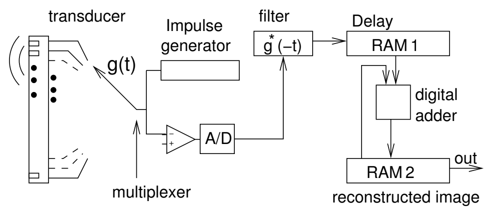
- Two types of reconstruction: (a) deconvolution and (b) matched filtering.
- Most often matched filtering. It can be performed either in frequency or in time domain.
- Conventional delay-and-sum beamforming is a good approximation.

The high-resolution image $\mathbf{H}(x, z)$ is formed as:

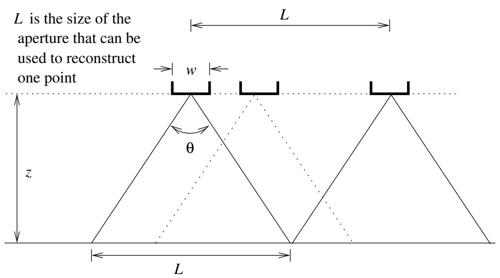
$$\mathbf{r}_i(t) = \mathrm{Re}\{r_i(t) \underset{t}{*} g^*(-t)\}$$

$$\mathbf{H}(x_p, z_p) = \sum_{i=1}^{N} \mathbf{r}_i(t_p(x_i))$$

$$t_p(x_i) = \frac{2\sqrt{(x_p - x_i)^2 + z_p^2}}{c}$$



# Influence of Finite Element Size




$$A_{t\ \mathrm{el}}(x) = \frac{\sin\left(\frac{k}{2} w \sin\theta\right)}{\frac{k}{2}\sin\theta}$$

$$\theta_a \approx 2 \arcsin \frac{\lambda}{2w}$$

Big elements reduce the size of the synthesized array.

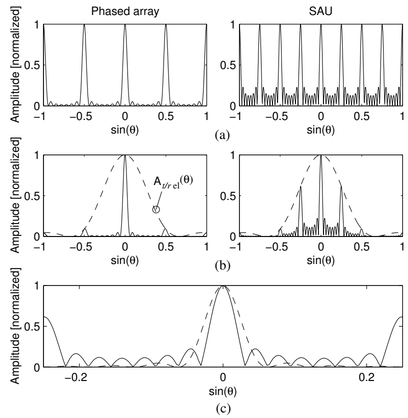


# Radiation Pattern - SAU vs Phased Array

$$A_{t/r\ \mathrm{SAU}\ \delta}(\theta) = \frac{\sin(kNd_x \sin\theta)}{\sin(kd_x \sin\theta)}$$

$$A_{t/r\ \mathrm{ph}\ \delta}(\theta) = \frac{\sin^2(\frac{k}{2} N d_x \sin\theta)}{\sin^2(\frac{k}{2} d_x \sin\theta)}$$

$$A_{t/r\ \mathrm{el}}(\theta) = \left(\frac{\sin(\frac{k}{2} w \sin\theta)}{\frac{k}{2}\sin\theta}\right)^2$$






# Monostatic SAU - Summary

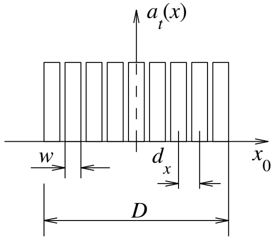
- The resolution of a SAU system is higher than of a phased array system.
- The resolution is depth independed if a sufficiently large array is available.
- The image is perfectly focused at *all* points in the image
- The side-lobe level of SAU is also higher
- The grating lobes appear closer to the main lobe for a SAU system.
  To push them outside the visible region, the pitch of the array must be $\lambda/2$.

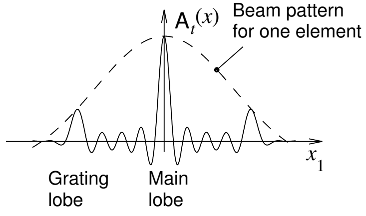


# Synthetic Transmit Aperture Imaging

- Preliminary concepts - Effective Aperture
- Image in one emission - Low Resolution Images (LRI)
- Geometrical model of LRI
- Examples
- Forming of High Resolution Images (HRI)
- LRIs and monostatic SAU imaging

# Effective Aperture



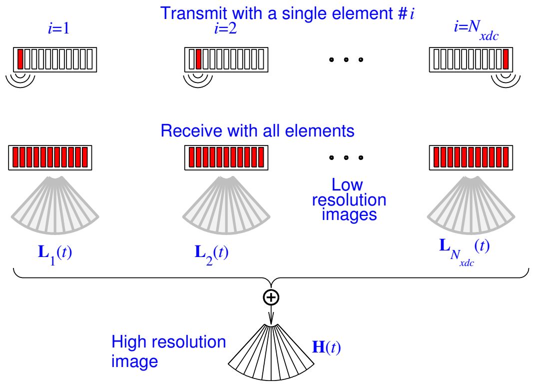

$$a_t(x_0) \rightleftharpoons \mathsf{A}_t(x_1)$$
$$a_r(x_0) \rightleftharpoons \mathsf{A}_r(x_1)$$
$$\mathsf{A}_{t/r}(x_1) = \mathsf{A}_t(x_1) \cdot \mathsf{A}_r(x_1)$$

$$a_{t/r}(x_0) \rightleftharpoons \mathsf{A}_{t/r}(x_1)$$
$$a_{t/r}(x_0) = a_t(x_0) \underset{x}{*} a_r(x_0)$$

$a_{t/r}$ - effective aperture



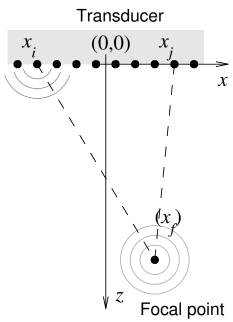
# Synthetic Transmit Aperture Imaging





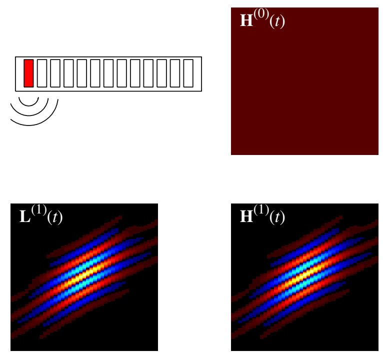

# STA - Beamforming

$$\mathbf{L}_i(\vec{x}) = \sum_{j=1}^{N_{xdc}} a_{ij}(\vec{x}) r_{ij}(t_{ij}(\vec{x}))$$
$$\mathbf{H}(\vec{x}) = \sum_{i=1}^{N_{xdc}} \mathbf{L}_i(\vec{x})$$
$$t_{ij}(\vec{x}) = \frac{|\vec{x} - \vec{x}_i| + |\vec{x} - \vec{x}_j|}{c}$$

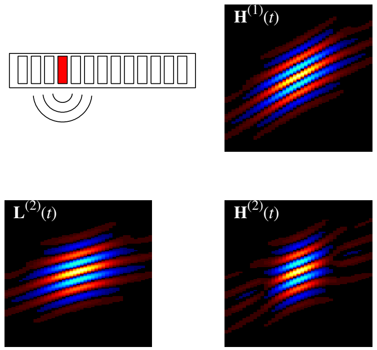





# STA Imaging (1)

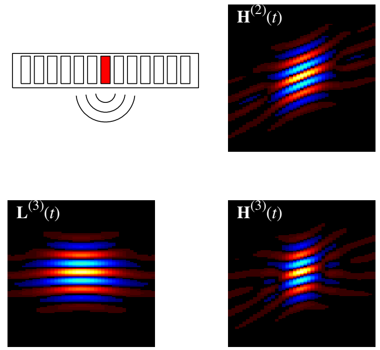


# STA Imaging (2)



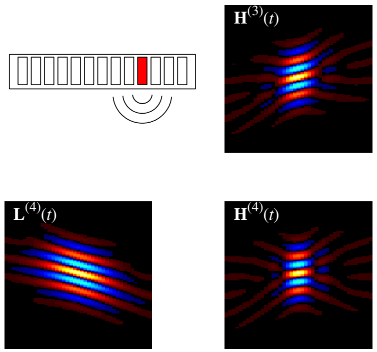



# STA Imaging (3)



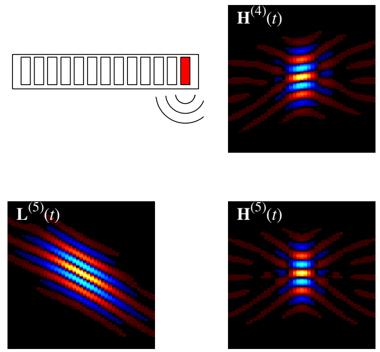



# STA Imaging (4)

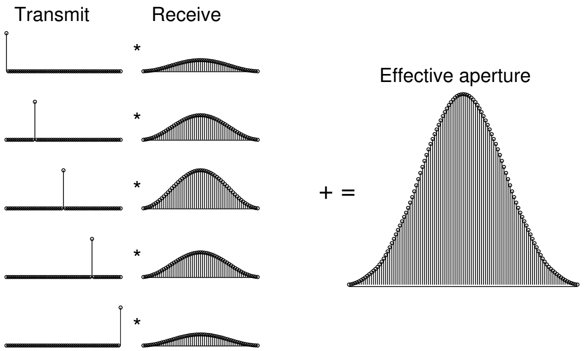





# STA Imaging (5)

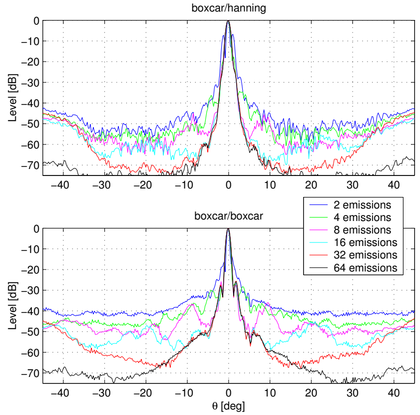


## Sparse Synthetic Transmit Aperture Imaging






## STA Imaging - Radiation Patterns






## Benefits From Synthetic Aperture Imaging

- SAU imaging gives excellent focusing
- SAU imaging is fast
- SAU is scalable to real-time 3D imaging

**Practical Problems**

- Improve the SNR
- Estimate the flow



# *Synthetic Aperture Ultrasound Systems*

## *Part 2: Implementation in Medical Ultrasound Imaging*

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# Outline

1. Implementation issues (overview)
2. Improving SNR
    - Solutions: Multi-element defocused apertures, temporal encoding
    - Imaging examples: Phantom & in-vivo
3. Tissue motion
    - Influence on image quality
    - Compensation
    - Imaging examples



# Practical Implementation: An overview

- Three known limitations to real-time in-vivo application are:
    1. Low SNR
    2. Tissue motion
    3. Real-time processing - dual stage beamforming

1. Low SNR
    - Occurs due to the single transmit element
    - Limits the penetration depth and thus in-vivo image quality

2. Tissue motion
    - Will be a problem if motion is severe
    - Results in incoherent summation of the low resolution images
    - Image smearing and loss in image quality



# Methods to Improving the Signal-to-Noise-Ratio

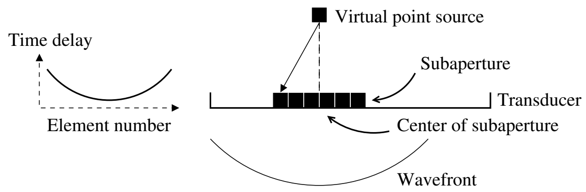
Known methods to increase the SNR of SA imaging:

- Spatial (aperture) encoding:
    1. Multi-element defocused subapertures
    2. Hadamard Encoding
       (e.g. Chiao et al. IEEE Ultrasonics Symp. 1997)
- Temporal encoding:
    1. Linear frequency modulated (FM) signals (chirp signals)
    2. Orthogonal complementary codes (Golay codes)
       (e.g. US Patent 6048315 and corresponding paper: Chiao and Thomas, IEEE Ultrasonics Symp. 2000)



# Multi-Element Defocusing

The purpose is to emulate the radiation pattern of a single high power transmit element, i.e. a virtual point source.



## Multi-Element Defocusing: Transmit Delays

- The transmit delays are derived using the geometric relations:

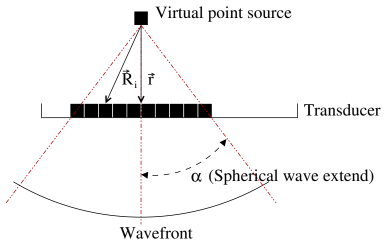
$$\tau(i) = \frac{|\vec{R}_i| - |\vec{r}|}{c}$$

where $c$ is the sound speed.

- The angular extend $\alpha$ of the spherical wavefront is determined by:

$$\tan(\alpha) = \frac{D}{2|\vec{r}|}$$

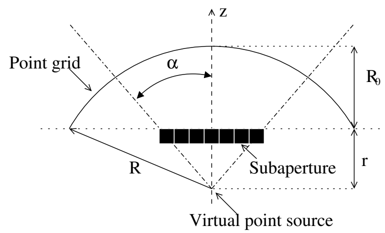
where $D$ is the size of the subaperture.





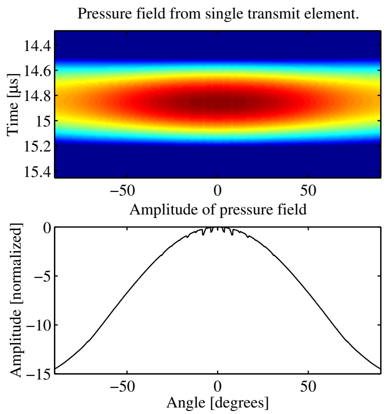

## Multi-Element Defocusing: Wavefront Calculation

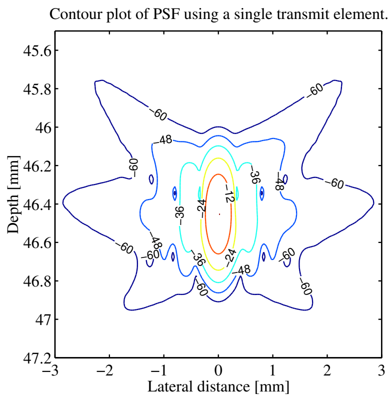
- Wavefront generation is complex when aperture is large
- Use Field II to calculate the wavefront.
- Note: Center of wave is at the virtual source center






## Wavefront Examples: Single Element Transmission

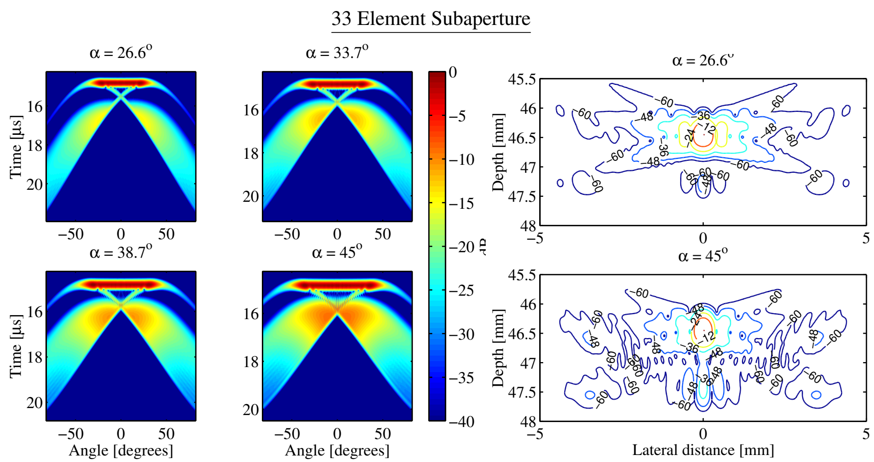



Perfect circular wavefront

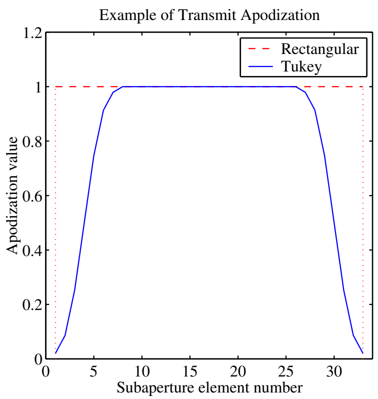


## Wavefront Examples: Multi-Element Subaperture



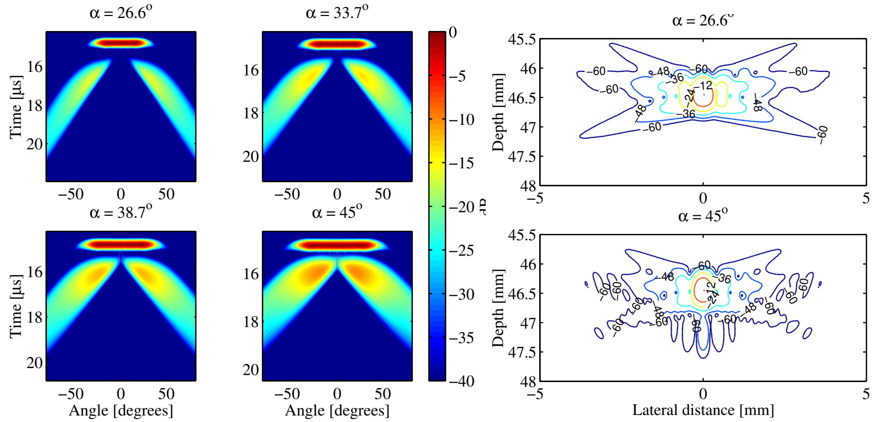
## Multi-Element Defocusing: Transmit Apodization

- Tails have a negative impact on the point spread function.
- Can be minimized to some degree by applying transmit apodization.
- Influence of edge waves can removed using e.g. a Tukey window



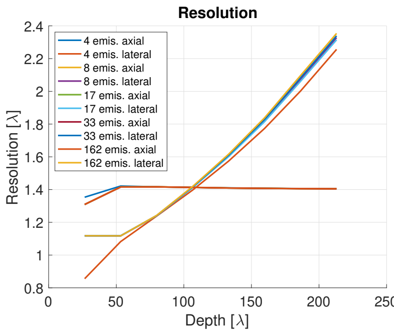


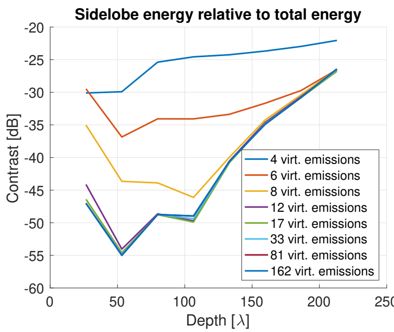

## Wavefront Examples: Tukey Apodization






## Resolution and contrast as a function of number of emissions

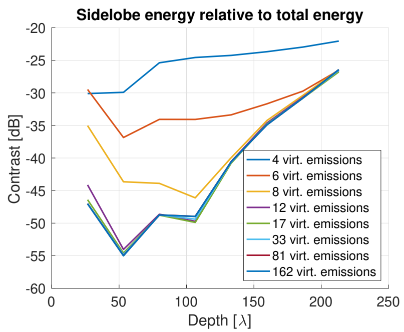


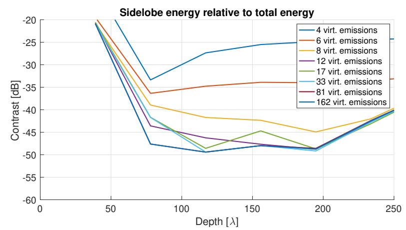




Linear array probe with $\lambda/2$ pitch and 192 elements. 32 elements used for emission



## Contrast as a function of number of emissions



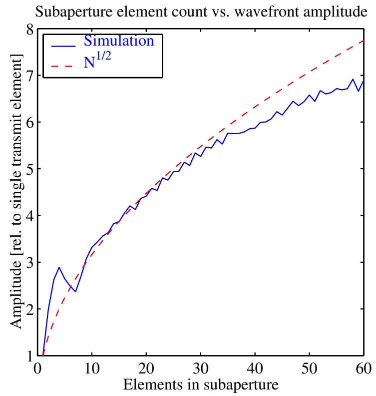




Linear array probe with 192 elements, 32 elements used for emission.

$\lambda/2$ pitch (left) $\lambda$ pitch (right).

## Wavefront Amplitude

Field II Simulation






## Multi-Element Defocusing: Beamforming Issues

- Limited spherical wave extend
  - Practically determines the sensitivity of the transmit source.
  - Puts a lower limit on the transmit f-number, i.e. lateral resolution.
- Take into account the virtual source locations.
  - The transmit aperture is defined by the spatial positions virtual sources.

In general, the sensitivity or acceptance angle of the elements needs to be taken into account when beamforming to optimize image quality.



## Methods to Improve the SNR: Status

Known methods to increase the SNR of SA imaging:

- Spatial (aperture) encoding:
  - *Multi-element defocused subapertures* $\surd$
- Temporal encoding:
  - Linear FM signals *NEXT*



## Temporal Encoding: Motivation

- (Peak) SNR after matched filtrering:

$$SNR = \frac{2E}{N_0}$$

  where $E$ is the signal energy, and $N_0$ is the white noise power spectral density [W/Hz].
- $E$ can be increased in two ways: *amplitude* or *duration*
- Cannot increase amplitude due to:
  - Front end limitations.
  - Acoustic output regulations.
- Long sinusoid not feasible due to loss in temporal resolution ($\delta t = \frac{1}{B}$)

”Need a long signal with short duration......(???)”

## Linear FM signals: Definition

- The linear FM signal is defined by:

$$s(t) = a(t)\cos(\phi(t)), \qquad |t| \leq \frac{T}{2}$$

with quadratic phase:

$$\phi(t) = 2\pi f_0 t + \pi \frac{B}{T} t^2$$

- Instantaneous frequency:

$$f(t) = \frac{1}{2\pi}\frac{d\phi(t)}{dt} = f_0 + \frac{B}{T}t$$

- Time-bandwidth product:

$$D = TB$$



## Linear FM signals: Compression Output

- The compression output is given by ($D \geq 20$):

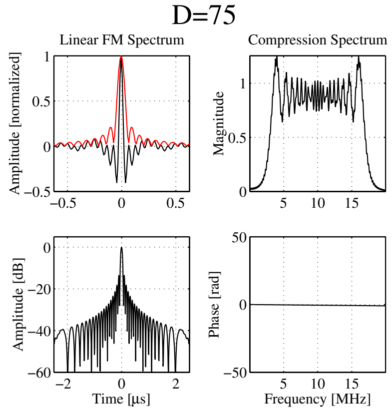
$$r(t) \simeq E \frac{\sin\left(\pi D \frac{t}{T}\right)}{\pi D \frac{t}{T}} \cos(\omega_0 t)$$

- Temporal resolution:

$$\delta t \simeq \frac{1}{B}$$

- Gain in SNR:

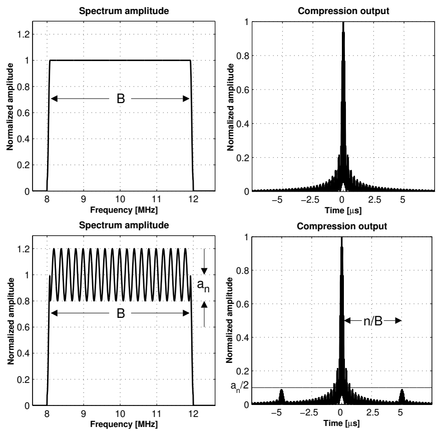
$$GSNR = \frac{E_{chirp}}{E_{pulse}} = \frac{T_{chirp}}{T_{pulse}}$$






## Linear FM signals: Sidelobe Characteristics

- The sharp edges produce near sidelobes
- The ripples in the pass-band produce distant sidelobes.
- Sharp rise time of the envelope produces pass-band ripples.

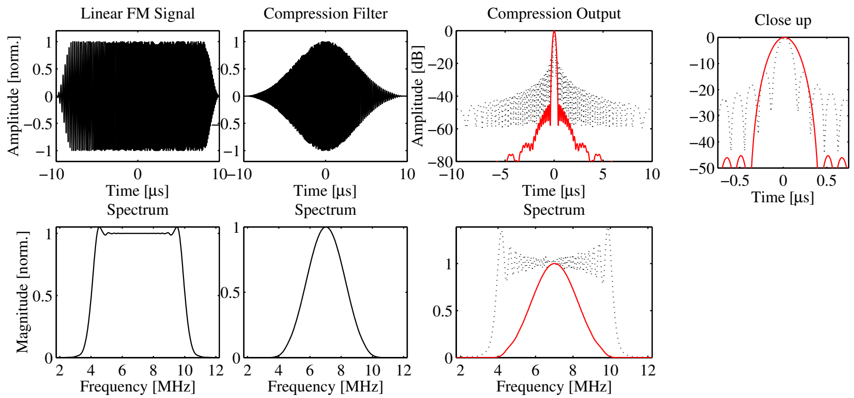



Sidelobes need to be reduced to obtain acceptable image contrast resolution!



## Linear FM signals: Sidelobe Reduction

- Edge weighting on excitation to suppress passband ripples.
- Smooth weighting on matched filter to minimize sharp spectrum edges



## SNR Model: Derivation

- Energy received by a single element from a point target placed at the acoustic focal point:
  - Conventional imaging: $E_{R_C} \sim N_{T_C}^2 T_C$
  - SA imaging: $E_{R_{SA}} \sim N_{T_{SA}} T_{SA}$
- (Peak) SNR after beamforming:
  - Conventional imaging: $SNR_C \sim N_{R_C} N_{T_C}^2 T_C$
  - SA imaging: $SNR_{SA} \sim M N_{R_{SA}} N_{T_{SA}} T_{SA}$

Variables: Conventional: $N_{R_C}$ - Number of receive elements, $N_{T_C}$ - Number of transmit elements, $T_C$ - Pulse duration

SA: $N_{R_{SA}}$ - Number of receive elements, $N_{T_{SA}}$ - Number of transmit elements, $T_{SA}$ - Pulse duration, $M$ - Number of emissions combined



## SNR Model: Derivation - continued

- Gain in SNR:

$$GSNR \sim \frac{MN_{T_{SA}}}{N_{T_C}^2} \frac{N_{R_{SA}}}{N_{R_C}} \frac{T_{SA}}{T_C}$$

Assumptions:

- White electronic noise uncorrelated between channels
- Quantization noise can be neglected by assuming close to full scale input to ADCs.
- Perfect phase alignment in beamformer ⇒ pure averaging
- No diffraction and attenuation effects included.



## SNR Model: Calculation Example

Linear array imaging:

- $N_{T_C} = 64$ elements
- 2 cycle sinusoid @ 7 MHz ($T_C \simeq 0.3\ \mu s$)
- $N_{R_C} = 128$ elements

TMS imaging:

- $N_{T_{SA}} = 33$ elements
- $M = 96$ elements
- $N_{R_{SA}} = 128$ elements
- $T_{SA} = 20\ \mu s$

$GSNR \simeq 17$ dB

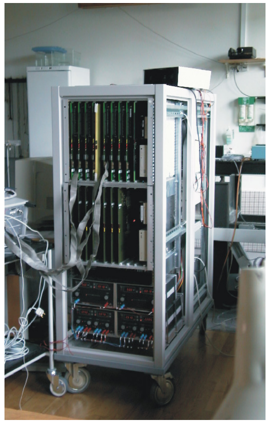


## Imaging Examples: Equipment - RASMUS

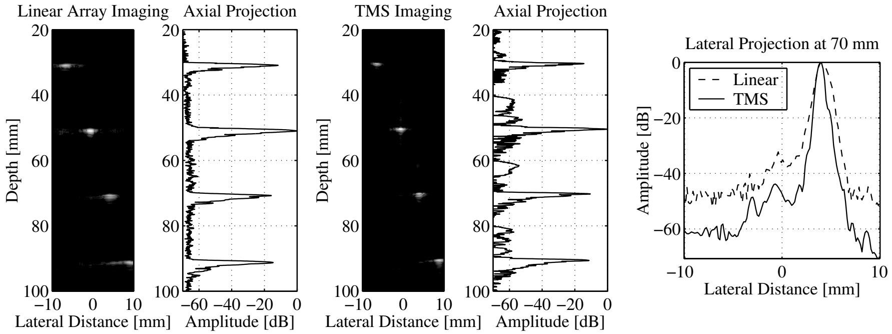
Show images using two transducer types: Linear array, and convex array

- Linear array:
  - 128 elements
  - $f_c = 7$ MHz
  - pitch = λ
  - $BW \simeq 50$ %
- Convex array:
  - 128 elements
  - $f_c = 5.5$ MHz
  - pitch = λ
  - $BW \simeq 60$ %

RASMUS



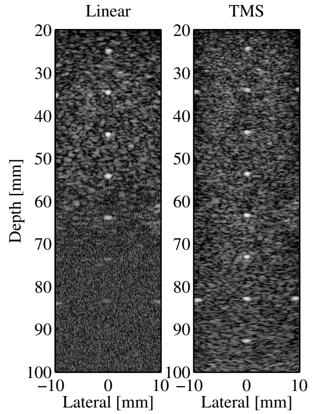
## Linear Array Transducer: Water Phantom

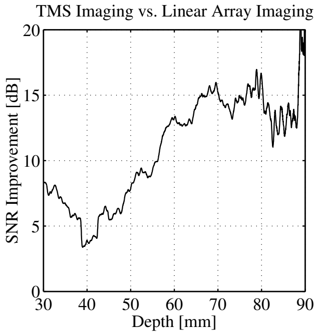



- Linear array imaging: $N_T = 64$, $N_R = 128$, 2 cycle sinusoid, dynamic receive beamforming.
- TMS imaging: $N_T = 33$, $N_R = 128$, $M = 96$, 20 $\mu s$ chirp, dynamic transmit and receive beamforming.



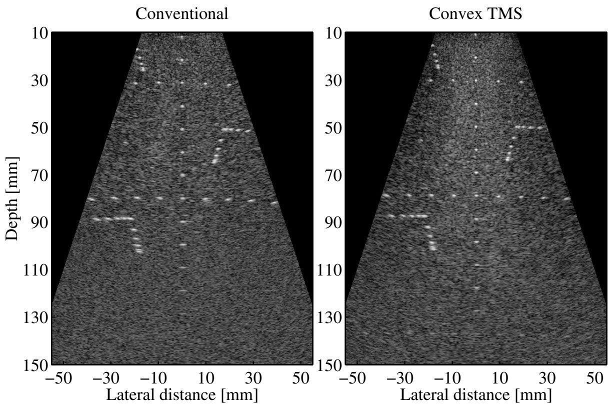
## Linear Array Transducer: Speckle Phantom





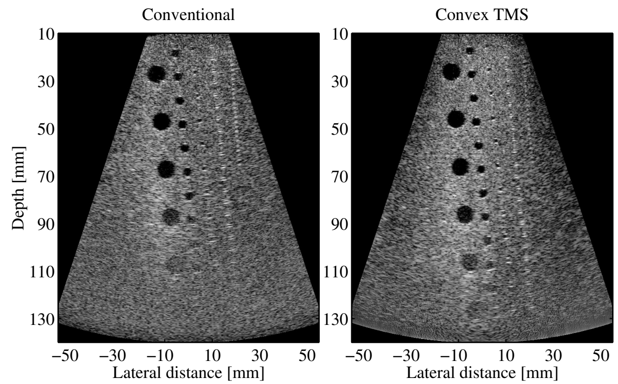

## Convex Array Transducer: Wire Phantom



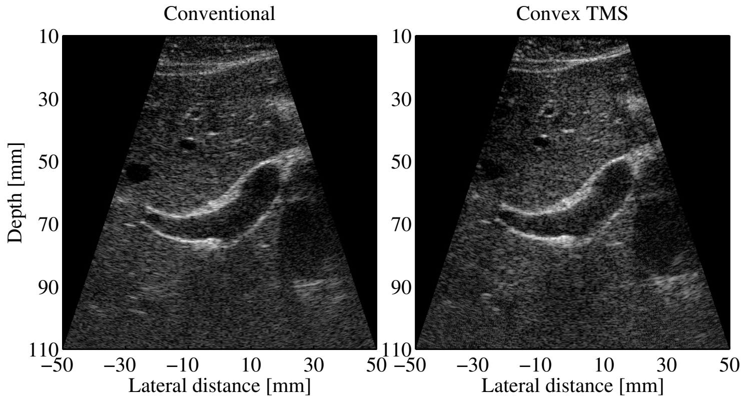

- Convex array imaging: $N_T = 64$, $N_R = 128$, 2 cycle sinusoid, dynamic receive beamforming.
- TMS imaging: $N_T = 11$, $N_R = 128$, $M = 118$, 20 $\mu s$ chirp, dynamic transmit and receive beamforming.



## Convex Array Transducer: Cyst Phantom



## Convex Array Transducer: In-vivo Abdominal






## Imaging Examples: Equipment - Verasonics scanner

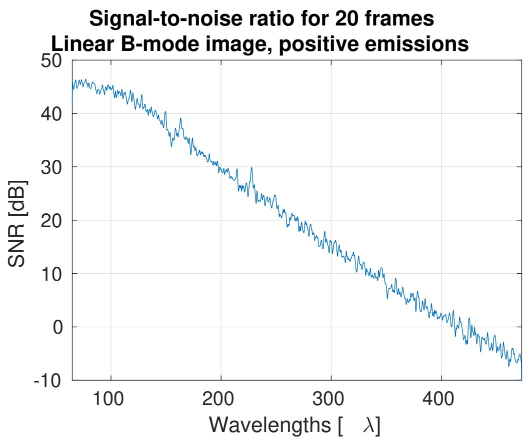
Commercial GE Hockey Stick probe GE L8-18i-D

- Linear array:
  - 168 elements
  - $f_c = 10$ MHz
  - pitch = λ
- SA sequence:
  - 12 virtual sources
  - $f_c = 10$ MHz, 1 cycle excitation
  - 32 elements in transmit, F# = -0.7
  - Dynamic focusing and apodization in receive
  - Cyst phantom with 0.5 dB/[MHz cm]

Verasonics research scanner



## SNR: Commercial GE Hockey Stick probe GE L8-18i-D




No codes. Pentration as for normal imaging.
Often normal transmission is sufficient.



## Overview

1. Implementation issues (overview) √
2. Improving SNR √
3. Tissue motion *NEXT*
   - Influence on image quality
   - Compensation
   - Imaging examples

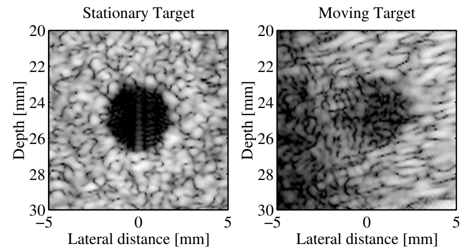
# Tissue Motion: Influence on Image Quality

- In conventional imaging, tissue motion causes image distortion.
  - A moving structure will not be presented in its natural shape
- In STA imaging, tissue motion produces (phase) shifts between the low resolution images.
  - This causes incoherent summation, and thus unfocused/smeared images.
  - The degradation in image quality will depend on several parameters, e.g. tissue velocity, direction of motion, acquisition speed.



# Example of Image Degradation

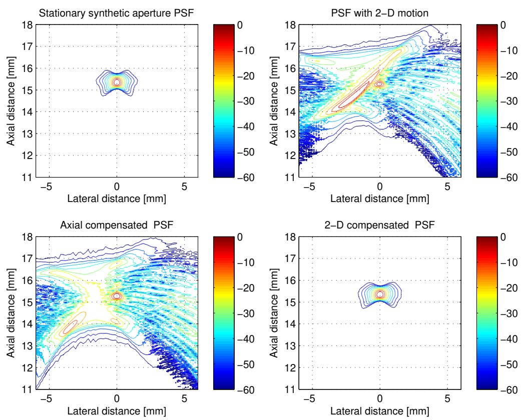
Field II Simulation




- Motion: 0.15 m/s at 70 degrees
- The contrast resolution has been degraded by 46.8%.



# Tissue Motion Compensation

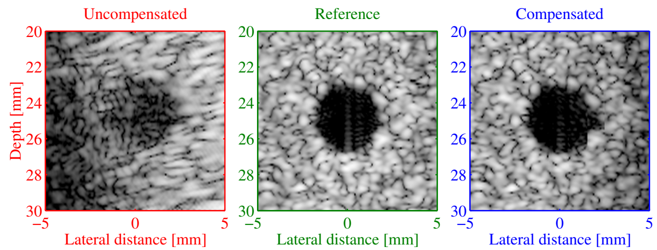



Made using SA vector flow imaging



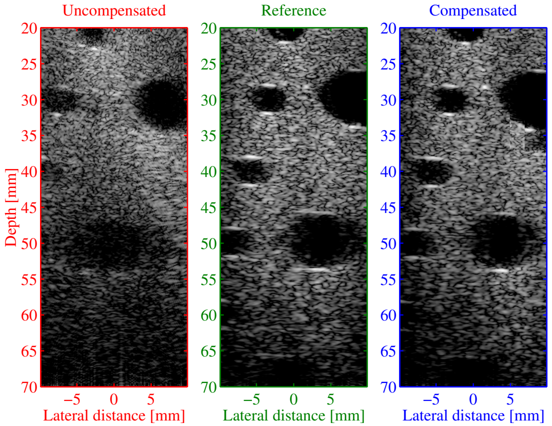
# Compensation Example: Simulation

Compensation using Directional Beamforming




- Motion: 0.15 m/s at 70 degrees
- The degradation in contrast before and after compensation is 46.8% and 0.1%, respectively

# Compensation Example: Measurement




- Directional beamforming method
- Uniform motion of 0.11 m/s at 68 degrees.
- The degradation in contrast before and after compensation is 55.3% and 1.6%, respectively.



# Synthetic Aperture Imaging Properties

- Frame rate can be made very high up to the pulse repetition frequency
- Imaging in the whole region continuously
- Dynamic focusing and apodization in transmit
- Optimal focusing and apodization in both transmit and receive
- SNR higher than in conventional imaging with codes
- SNR comparable to conventional imaging without codes
- MI and $I_{spta}$ much lower than in conventional imaging for unfocused emissions