超音波影像實驗室
Ultrasonic Imaging Laboratory

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Post-doc: 1

PhD Student: 8

Master Student: 5

Research Assistant: 2

Administration Assistant: 1

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PERSONAL EDUCATION

National Taiwan University	1983-1987	B.S.	Electrical Engineering
University of Michigan	1989-1990	M.S.	EE: Systems
University of Michigan	1991-1994	Ph.D.	EE: Systems

RESEARCH INTEREST

Signal and imaging processing, Ultrasonic medical imaging

WORK EXPERIENCE

Adjunct Associate Investigator	Health Research Institutes, Taiwan	2001-present.
Member of Technical Staff	Acuson Corporation, U.S.A.	1994-1997.

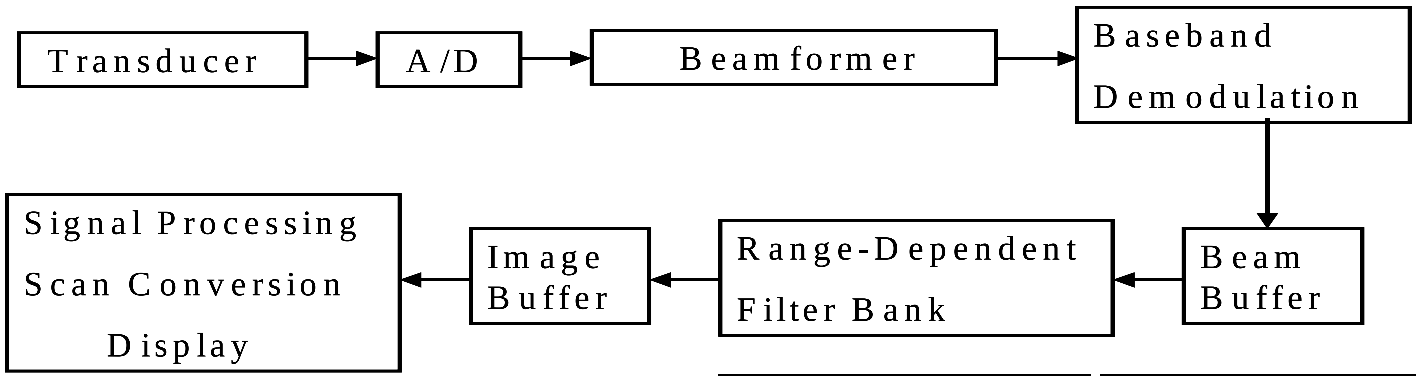
Present Programs Overview

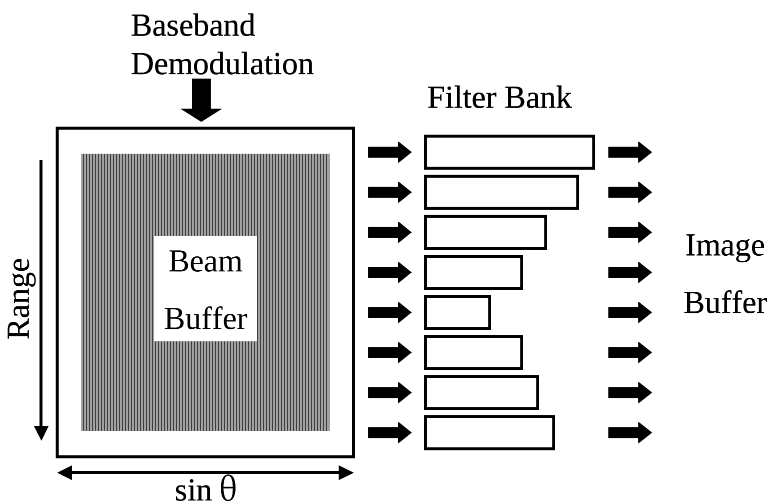
1. Advanced Beamforming Technology
2. Ultrasonic Small Animal Imaging
3. Ultrasound Assisted Liposomal Therapy
4. Opto-Acoustic Imaging
5. Others

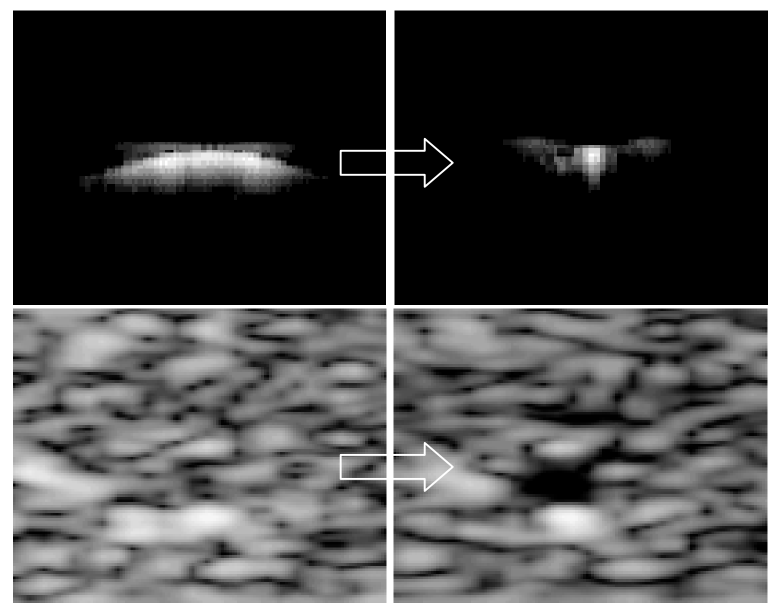
Advanced Beamforming Technology

Ultrasonic Synthetic Aperture Imaging

– Filter based synthetic focusing technique





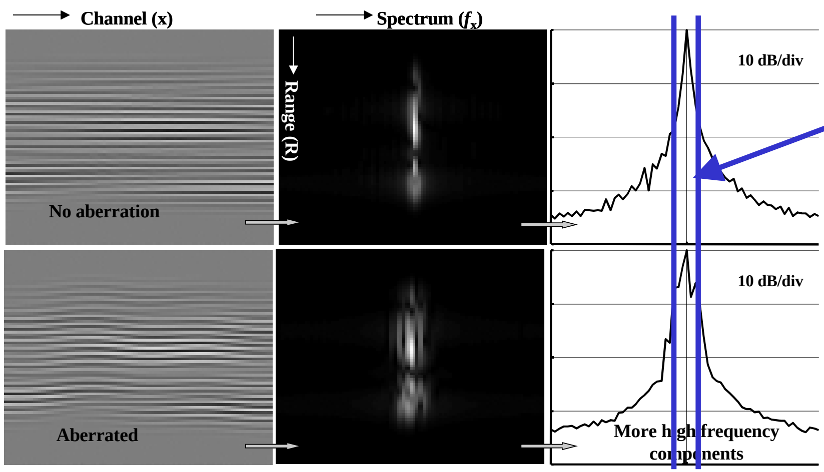


M.-L. Li and P.-C. Li, "Filter Based Synthetic Transmit and Receive Focusing", Ultrasonic Imaging, Vol. 23, pp. 73-89, April, 2001.

Adaptive Imaging

- A generalized-coherence-factor (GCF) weighting technique is proposed.

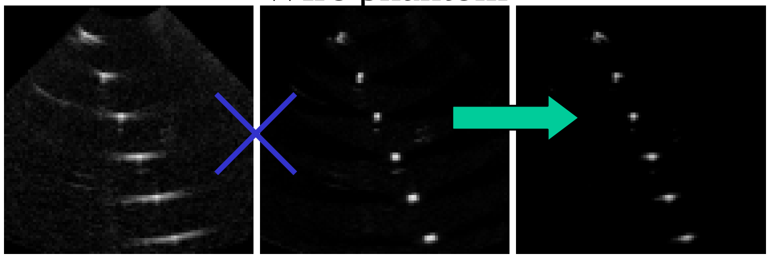
$$GCF = \frac{\text{spectral energy within a pre-specified low frequency range}}{\text{total spectral energy}}$$



High GCF corresponds to good focusing quality and the image intensity should be maintained

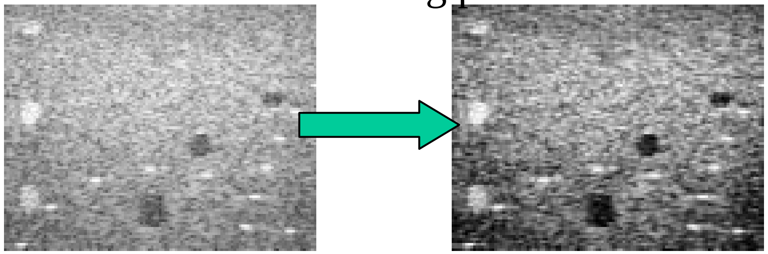
Lower GCF should be used to reduce the image data because significant beamforming errors are present

Wire phantom



Original GCF map GCF-weighted

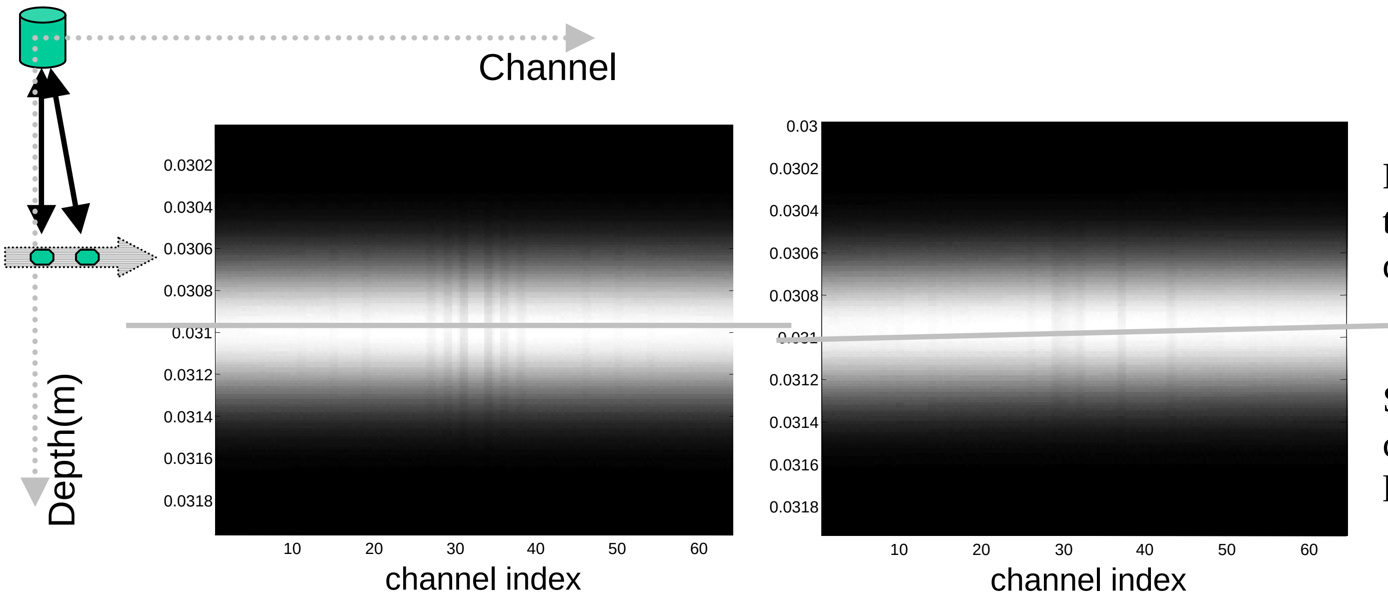
Tissue-mimicking phantom



Aberrated GCF-weighted

P.-C. Li and M.-L. Li, "Adaptive Imaging Using the Generalized Coherence Factor", IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control, Vol. 50, No.2, pp. 128-141, February, 2003.

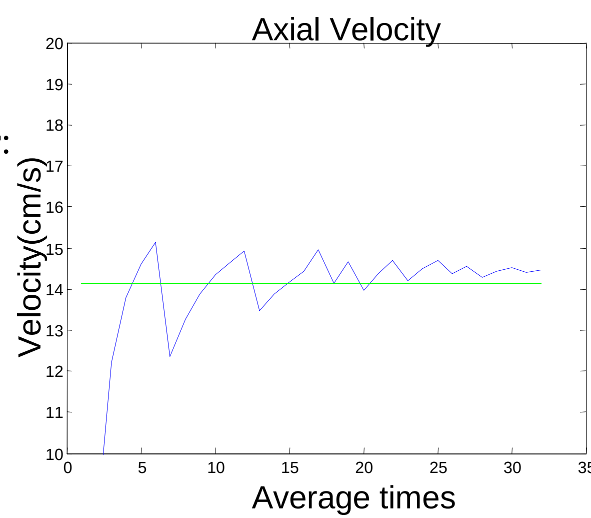
2-D Flow Estimation Using Channel Data

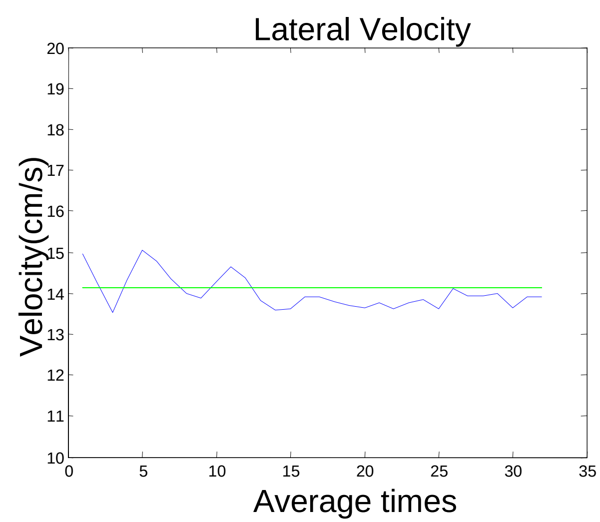


Lateral motion causes different tilt slopes in acquired channel data at different time.

Slope change in sequential channel data is related to lateral velocity.

Simulation results for only one scatterer:
Flow velocity=20cm/s,
Doppler angle=45^o,
SNR=10dB.

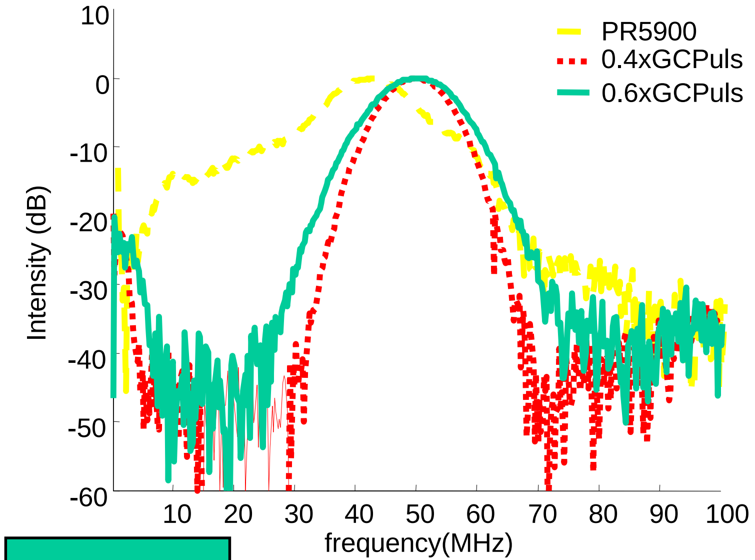


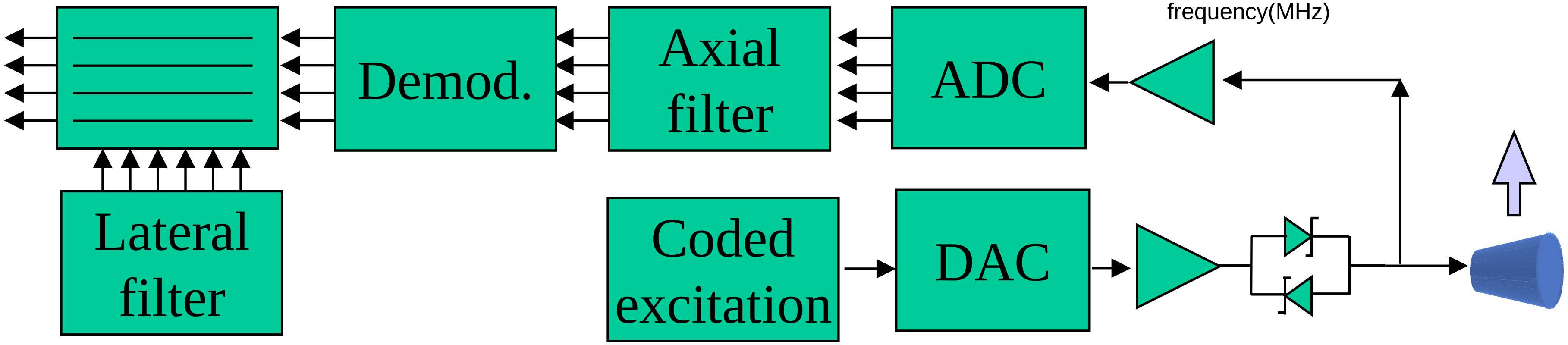


Ultrasonic Small Animal Imaging

High Frequency Ultrasonic Imaging System

- Fully digital system architecture
- 50 MHz center frequency
- 60 % fractional bandwidth



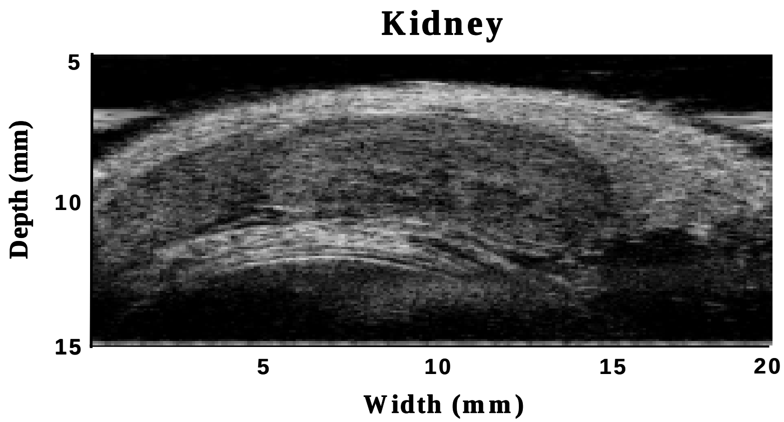


Dynamic focus

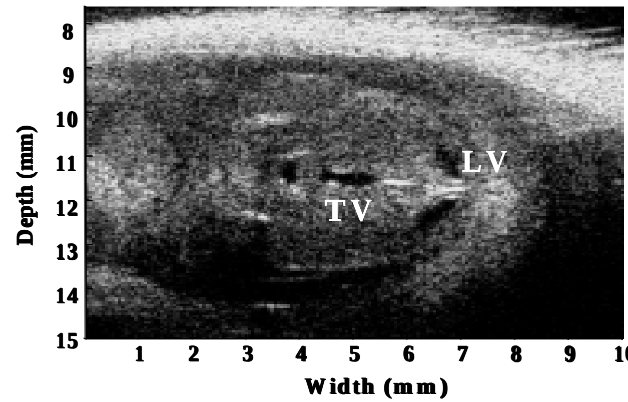
Tissue attenuation

Hardware Design

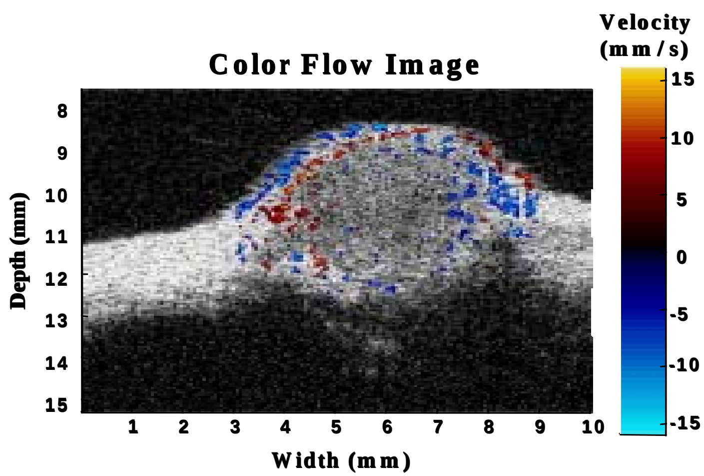
High Frequency Ultrasonic System: Rat

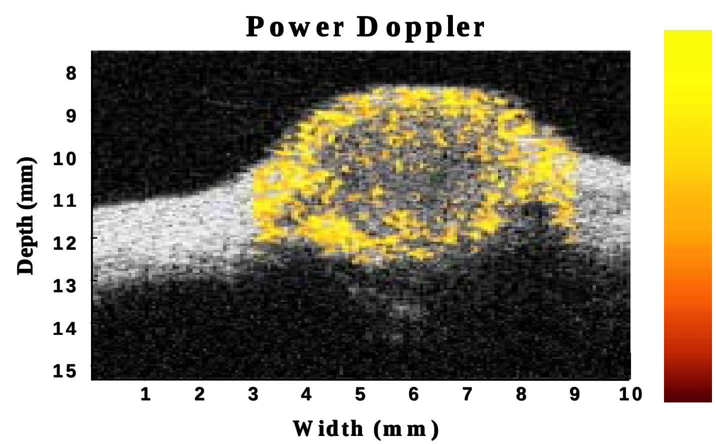


Rat B-mode kidney image



The B-mode images of embryo’s head.

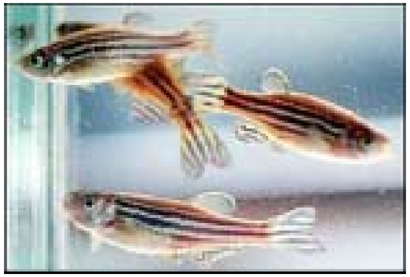




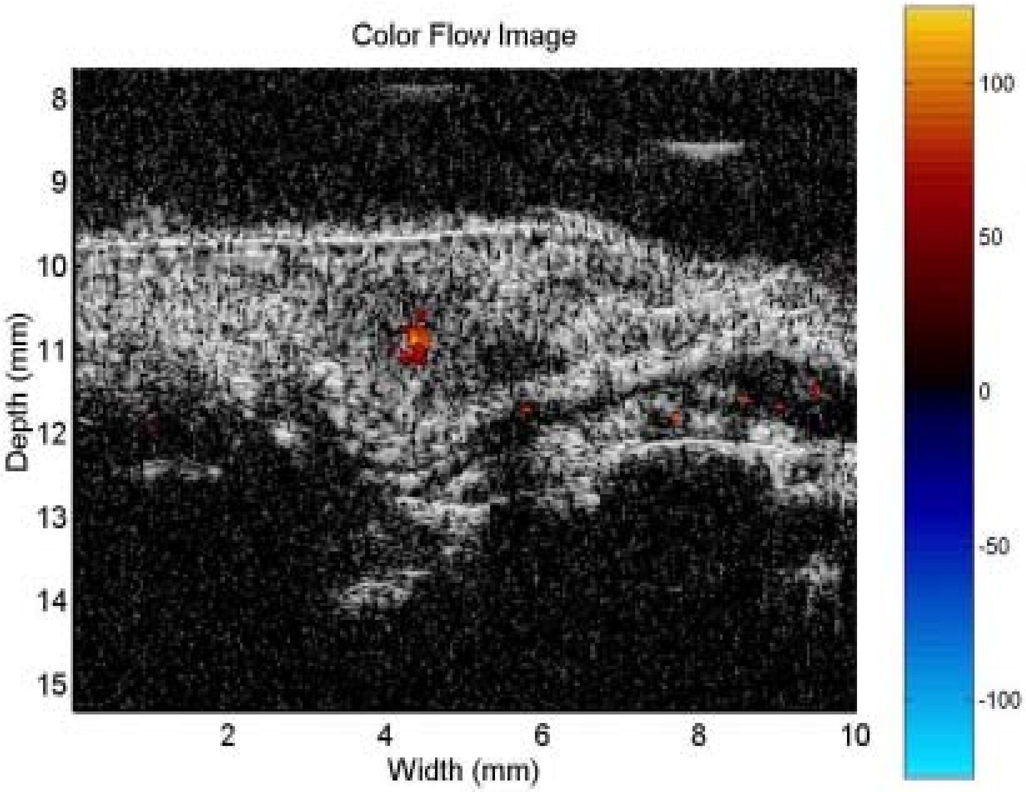
Rat Abdomen Tumor: Color Flow Image and Power Doppler.

High Frequency Ultrasonic System: Zebra-Fish

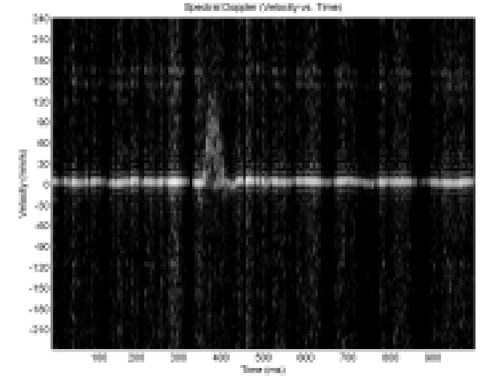
Zebra-Fish (3-5 cm)



Color flow image of Zebra-Fish in 25 MHz



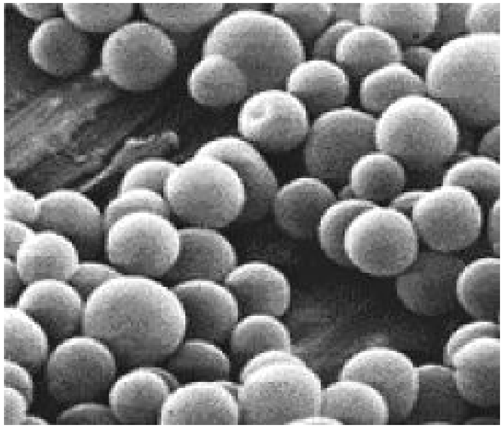
Spectral Doppler of heart



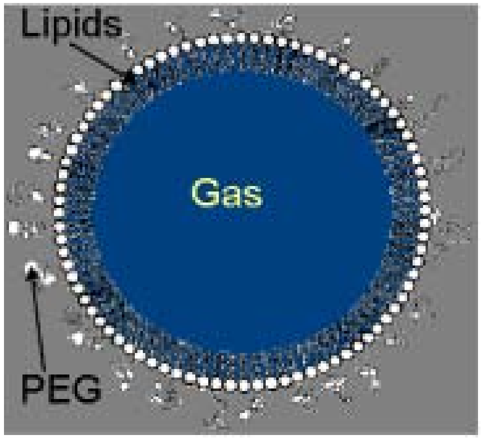
Ultrasound Assisted Liposomal Therapy

- **Ultrasound contrast agents (UCAs) are shell-encapsulated microbubbles**
- **UCAs are used to enhance backscattered echoes from blood (15-20 dB)**

contrast agents

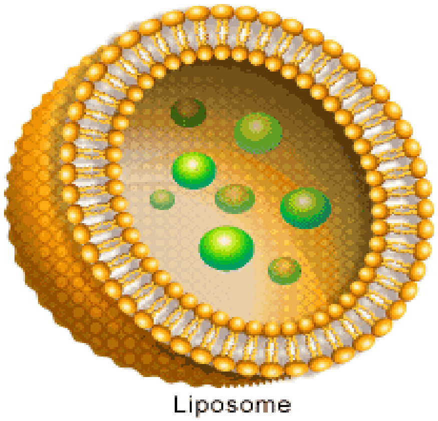


Contrast Agents (micorn)

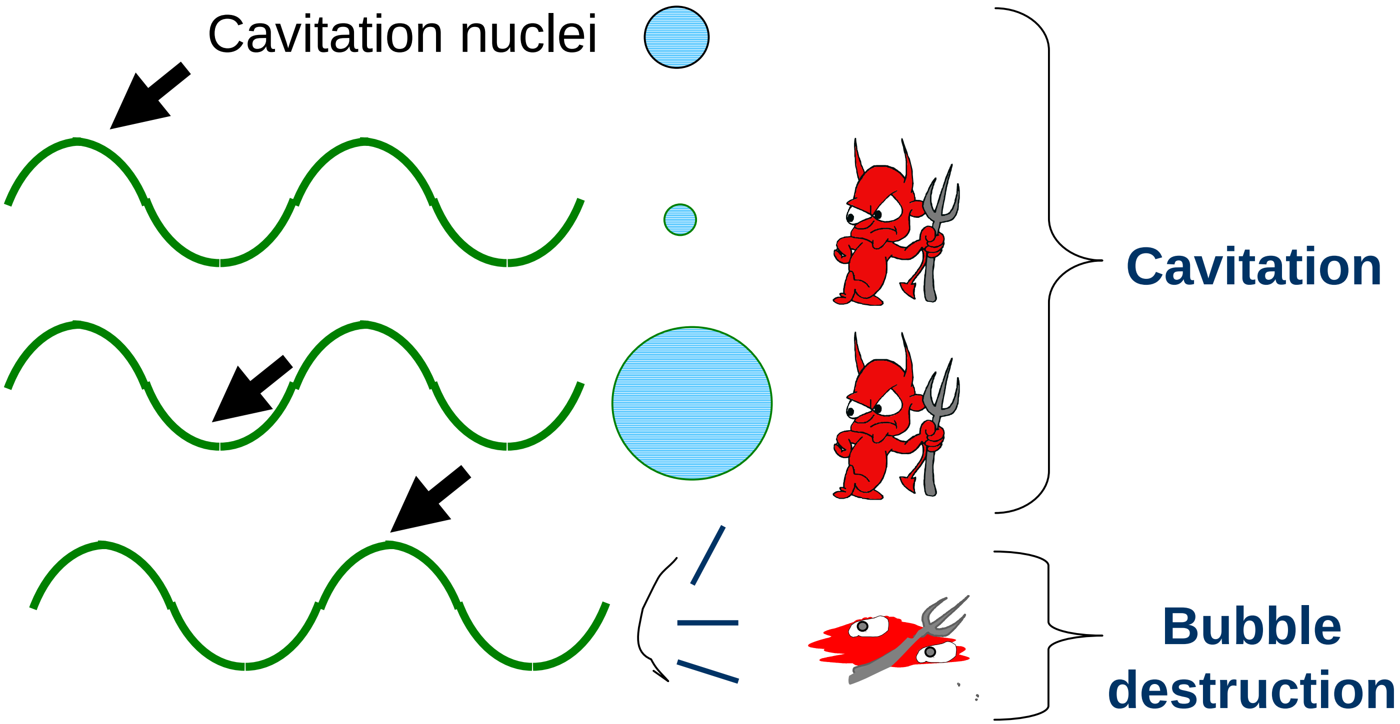


Not a drug vehicle

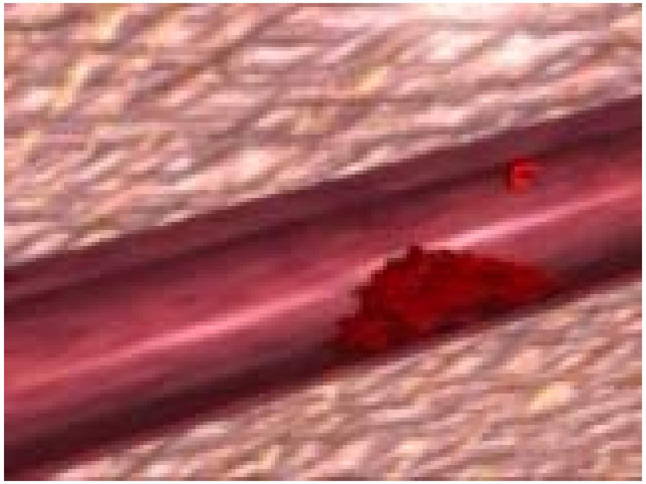
Liposomes (nano)



Mainly hydrophilic drug vehicle

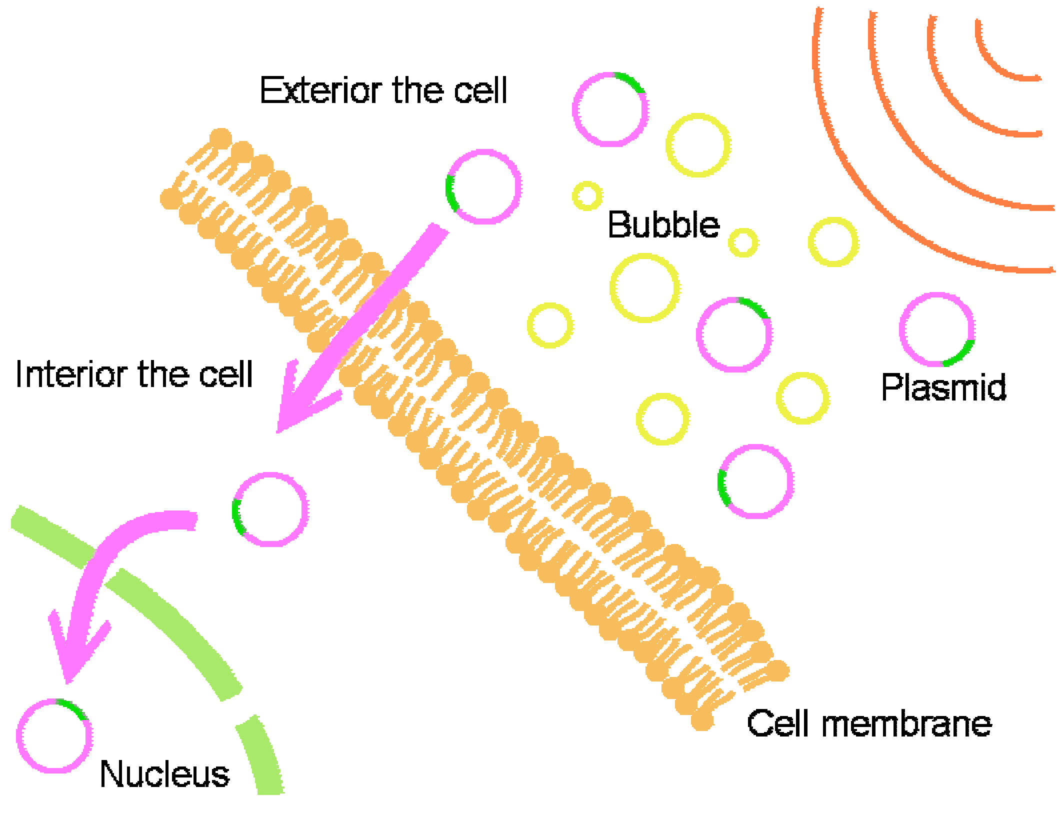


thrombus



Ultrasound on Gene Therapy

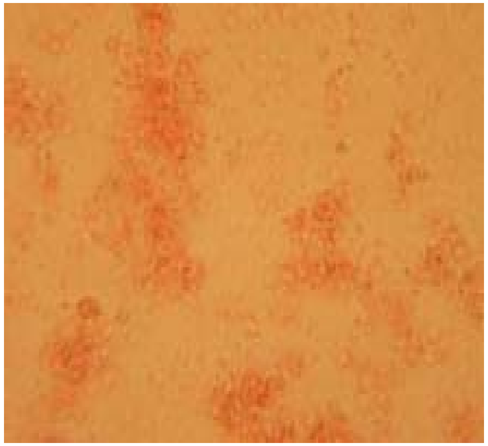
- Using Ultrasound, microbubbles, liposome and plasmid DNA to achieve gene transfer and gene expression
- Advantages :
  - Non-invasive
  - Target gene therapy
- Present works :
  - Mechanism research (cavitation, etc.)
  - Tumor inhibition and therapy



Opto-Acoustic Imaging

Opto-Acoustic Measurement of Blood Flow and Contrast Agent Fabrication

- Functional imaging. Ex : blood oxygen measurement.
- Develop O.A. contrast agent to enhance the O.A. signal.

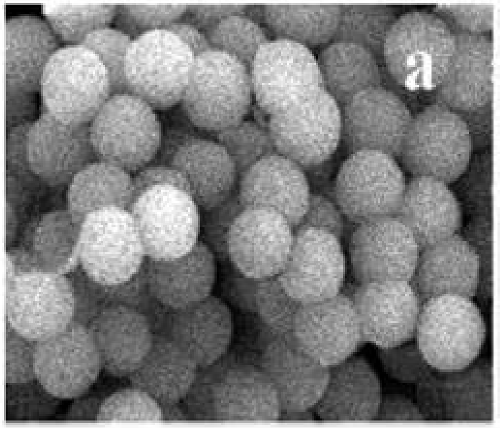


Liposome with dye
(Direct-81 red)

Current work:

1.Liposome (with dye Direct-81 red) as O.A. contrast gent

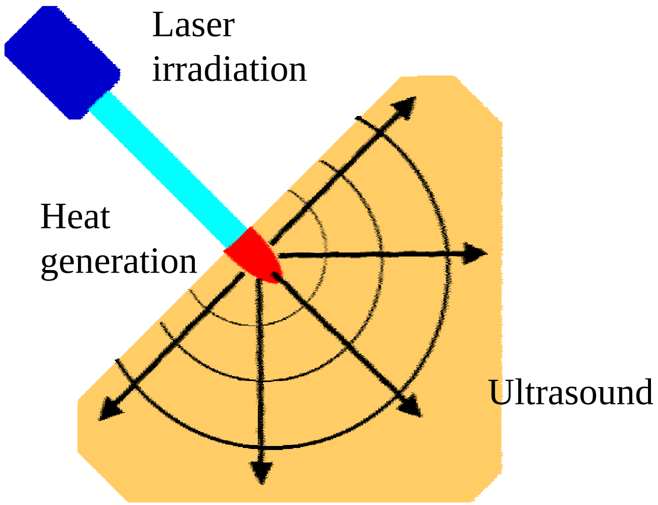
2.Gold nanoparticles as P.A. contrast agent

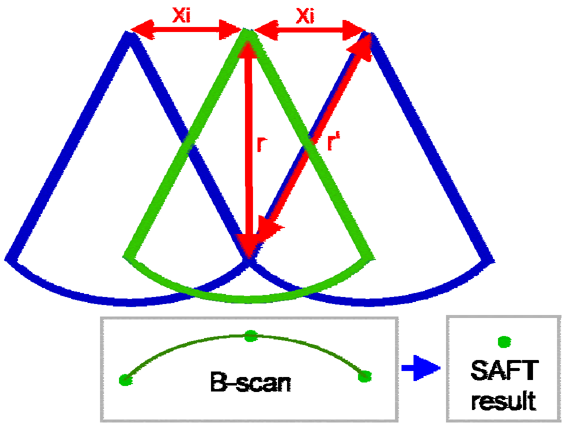


Gold nanoparticles

Improved Backward Opto-Acoustic Imaging Using Synthetic Aperture Focusing and Coherence Factor

- OA imaging was based on the different optical absorption coefficients in tissue.
- OA imaging has poor lateral resolution and SNR due to the wide optical and acoustic radiation patterns.
- Backward OA 2-D imaging system has been built up.
- Both the lateral resolution and SNR were improved by using SAFT and CF weighting method.

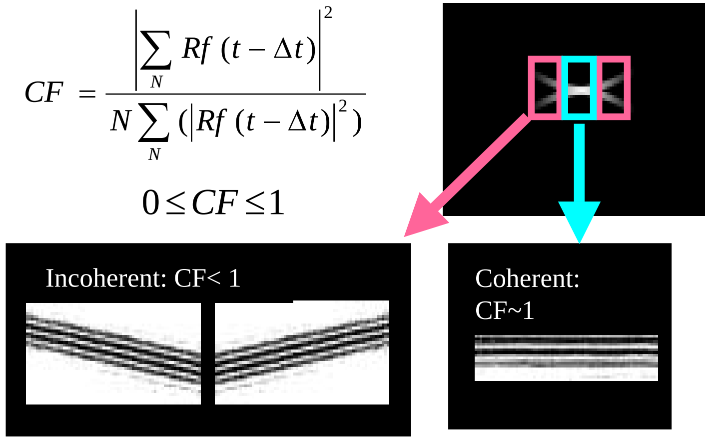




Synthetic Aperture Focusing Technique

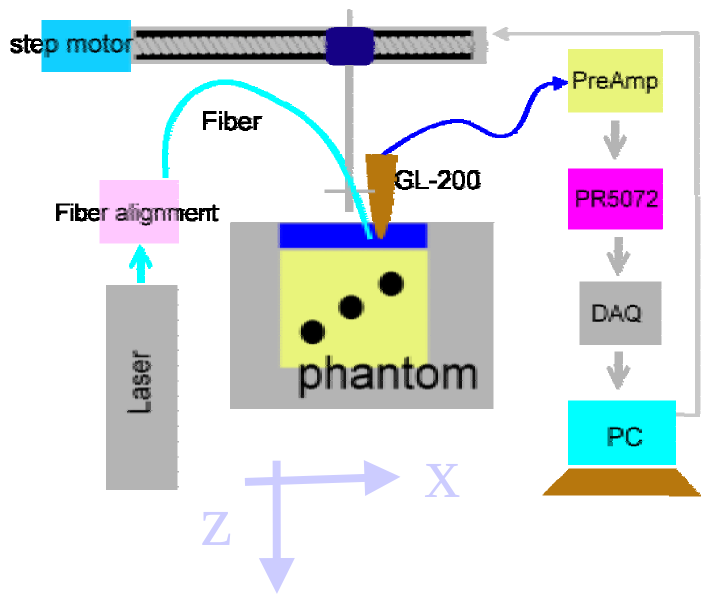
$$CF = \frac{\left|\sum_{N} Rf\,(t-\Delta t)\right|^2}{N\sum_{N}\left(\left|Rf\,(t-\Delta t)\right|^2\right)}$$

$$0 \le CF \le 1$$

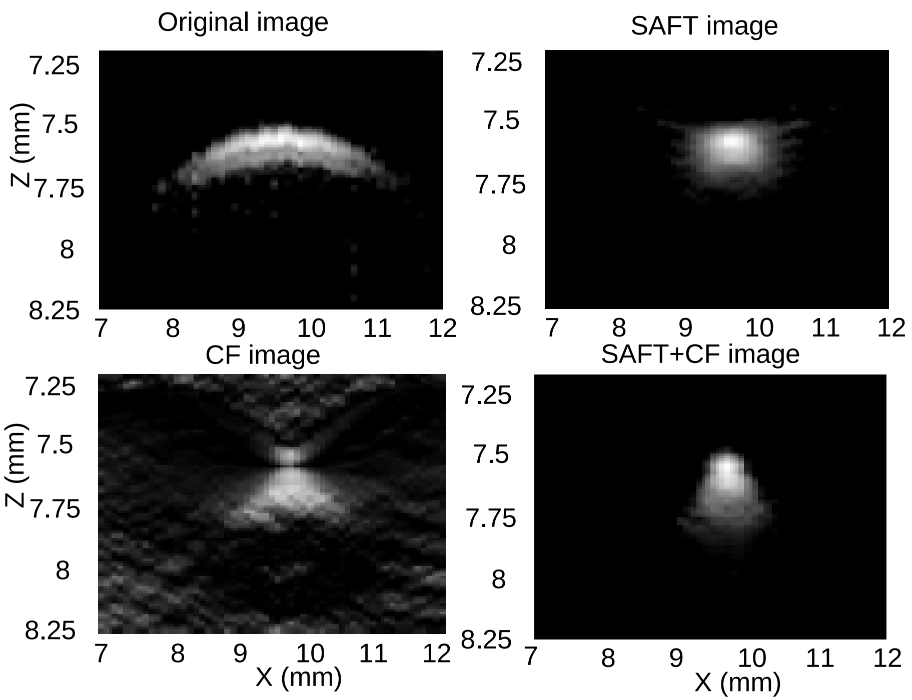


Coherent factor diagram

Cont'd



The OA B-mode image scan system



The original OA imaging of the hair phantom in milk medium (top left), the image after SAFT calculation (top right), the CF map (bottom left), the image after SAFT and CF weighting

A Numerical Approach for Opto-Acoustic Ultrasound

- Governing Equations

$$\frac{\partial \rho'}{\partial t} + \rho_0 \nabla \cdot \vec{u} = -\rho' \nabla \cdot \vec{u} - \vec{u} \cdot \nabla \left(\rho' + \rho_0 \right)$$

$$\rho_0 \frac{\partial \vec{u}}{\partial t} + \nabla p = -\rho' \frac{\partial \vec{u}}{\partial t} - \left(\rho_0 + \rho' \right) \left(\vec{u} \nabla \right) \vec{u} + \frac{1}{3} \mu \nabla \left(\nabla \cdot \vec{u} \right) + \mu \nabla^2 \vec{u}$$

$$\rho_0 T_0 \frac{\partial s}{\partial t} = \nabla \cdot \left(\kappa \nabla T' \right) + W$$

$$p - c_0^2 \rho' = \frac{c_0^2}{\rho_0} \frac{B}{2A} \rho'^2 + c_0^2 \left(\frac{\rho \beta T}{c_p} \right)_0 s$$

$$T' - \left(\frac{T \beta}{\rho C_p} \right)_0 p = \left(\frac{T}{C_v} \right)_0 s$$

ρ' : density deviation

$\vec{u}$: particle velocity

T' : temperature deviation

s : entropy deviation $\left({}^{J}\!/_{kg \cdot {}^{o}K} \right)$

p : acoustic pressure

W : heat generation function $\left({}^{W}\!/_{m^3} \right)$

κ : thermal conductivity $\left({}^{W}\!/_{m \cdot {}^{o}K} \right)$

μ : viscousity $\left({}^{kg}\!/_{m \cdot s} \right)$

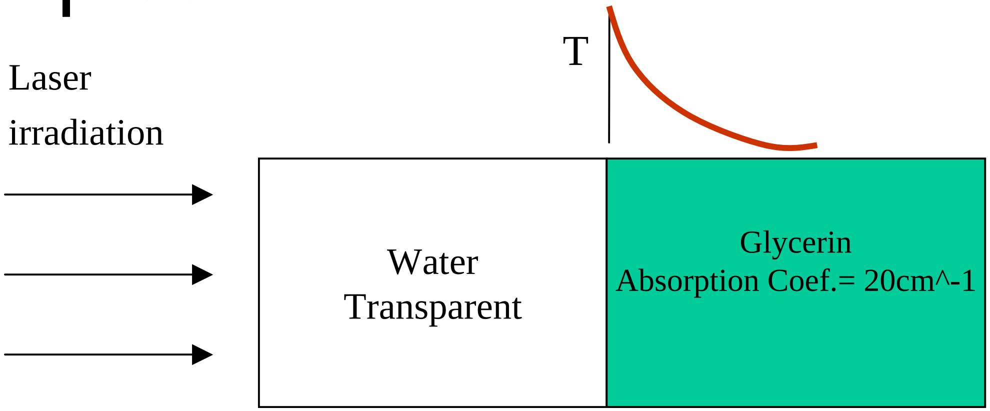
C_p, C_v : specific heats $\left({}^{J}\!/_{kg \cdot {}^{o}K} \right)$

ρ_0, T_0 : ambient density & temperature distribution

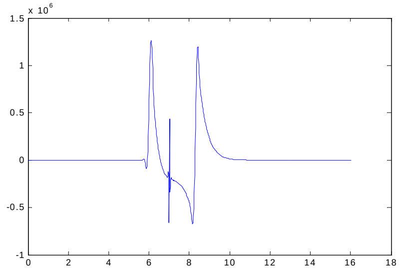
- All spatial inhomogeneities are taken into consideration.
- Given the initial values of $\rho', \vec{u}, s$, all variables at the future time can be found using the finite-difference time-domain (FDTD) method

Cont’d

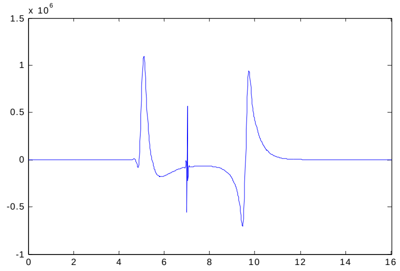
- Examples



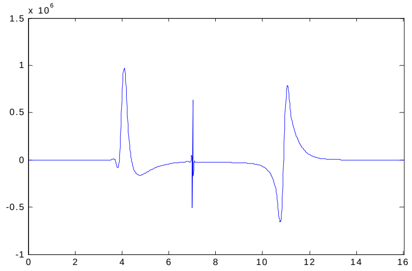
OA waveforms at selected moments



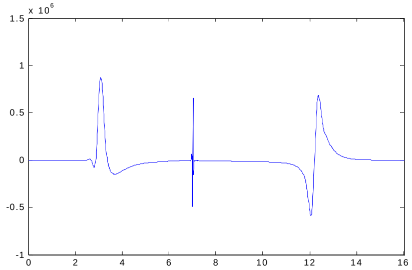
t = 0.066 us



t = 0.132 us



t = 0.198 us



t = 0.264 us

Others Studies

Pspice Modeling of Ultrasound Transducer

- **Model of an ultrasound single element transducer.**

- **Low acoustic impedance, low acoustic quality factor and low dielectric constant of piezoelectric polymer is suitable to fabricate ultrasound transducer.**

- **Acoustic and electrical part of transducer can easily be varied and analyzed by using Pspice simulation.**

- **In future, transducer model can expand to a complete ultrasound system**

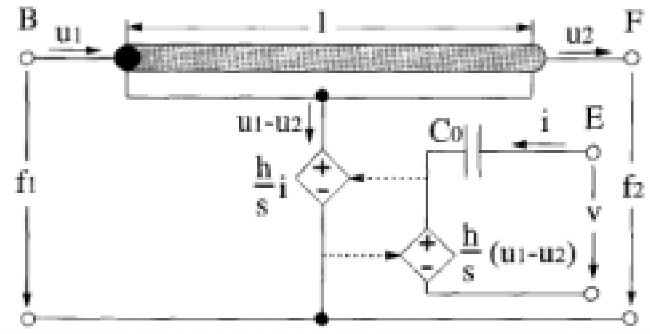


Fig. 1. Equivalent circuit for the thickness mode transducer

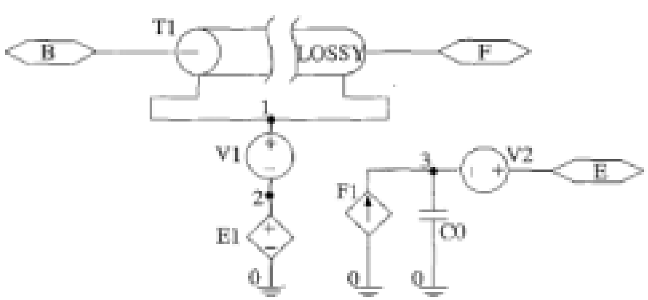
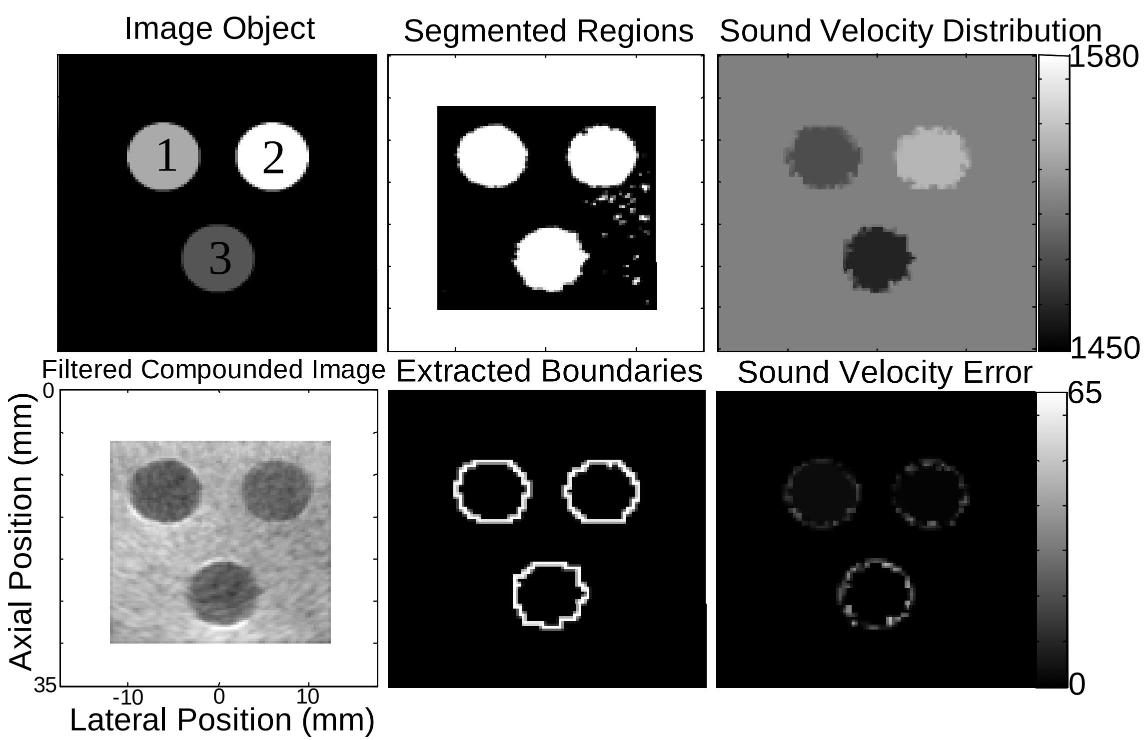


Fig. 2. Pspice subcircuit for thickness mode transducer

Computed Tomography Sound Velocity Reconstruction

- We proposed a method for incorporating the segmentation information of a B-mode image into the process of sound velocity reconstruction with limited-angle transmission tomography.
- The reconstructed sound velocities are accurate except at the boundaries.
- The sound velocity error are generally 1–3 m/s.
- Obtaining the sound velocity distribution is feasible with current B-mode imaging setup using linear arrays.

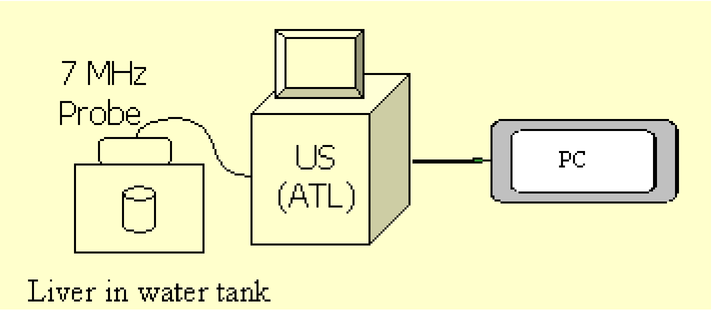


Combining High Frequency Ultrasound and Micro-PET System for Small Animal Imaging Study

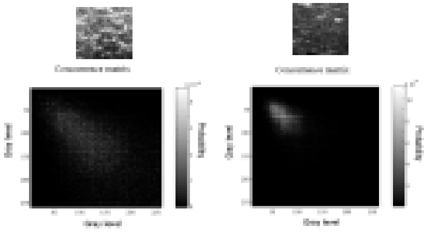
- ☐ small animal model and tumor monitoring
- ☐ using high frequency ultrasound system to measure tumor growth curve and angiogenesis
- ☐ micro-PET study :
  1. principle of tumor [18F]FDG PET detecting
  2. principle of PET imaging
  3. micro-PET imaging and
- ☐ radiopharmacokinetic study registration、 imaging fusion and to integrate the two imaging systems

Liver Fibrosis Grade Classification Using B-mode Ultrasound
以二維超音波影像特徵作肝纖維化程度之分類

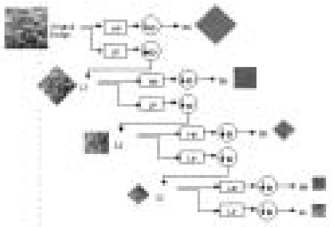
- Experiment set-up



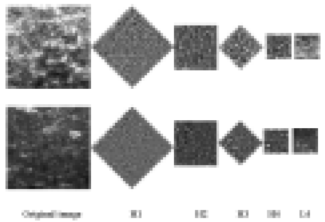
- Image feature extracted by gray level concurrence and non-separable wavelet transform
  - GLC (the energy of concurrence matrix of healthy liver is more concentrated then cirrhotic liver)



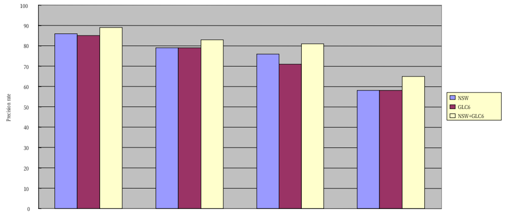
  - NSW



- Cirrhotic liver
- Healthy liver



  - The accuracy of different classes done by support vector machine



Previous Studies Overview

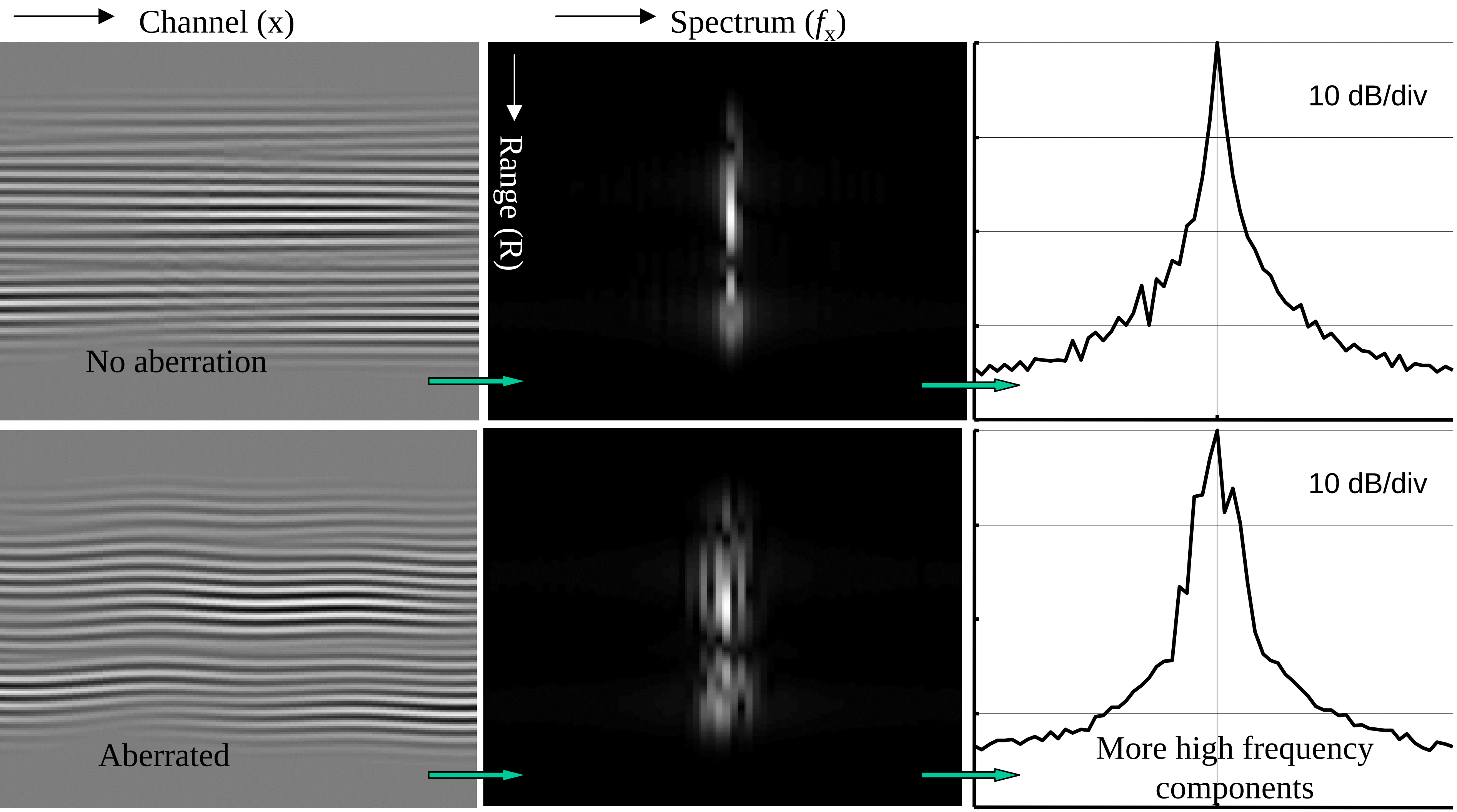
1. Adaptive Imaging
2. Ultrasonic Nonlinear Imaging
3. Ultrasonic Elastic Imaging
4. 3-D Ultrasonic Imaging
5. High Frequency Ultrasonic Imaging
6. Blood Flow Estimation Using Ultrasonic Contrast Agent

1. Adaptive Imaging

Adaptive Imaging

- A new adaptive imaging technique using generalized coherence factor (GCF) is proposed.
- GCF is derived based on the spectrum of the received array data along the array direction.
- GCF is an index of beamforming quality
- GCF is used as a weighting factor to the reconstructed image.

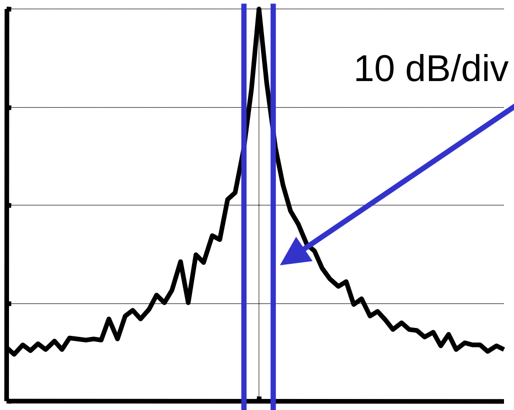
Idea



Channel (x)
Spectrum (f_x)
Range (R)
10 dB/div
No aberration
10 dB/div
Aberrated
More high frequency components

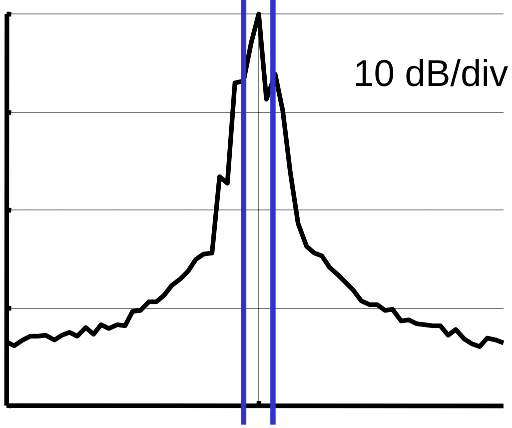
Generalized Coherence Factor

$$\mathrm{GCF} = \frac{\text{spectral energy within a pre-specified low frequency range}}{\text{total spectral energy}}$$



- High GCF corresponds to good focusing quality and the image intensity should be maintained

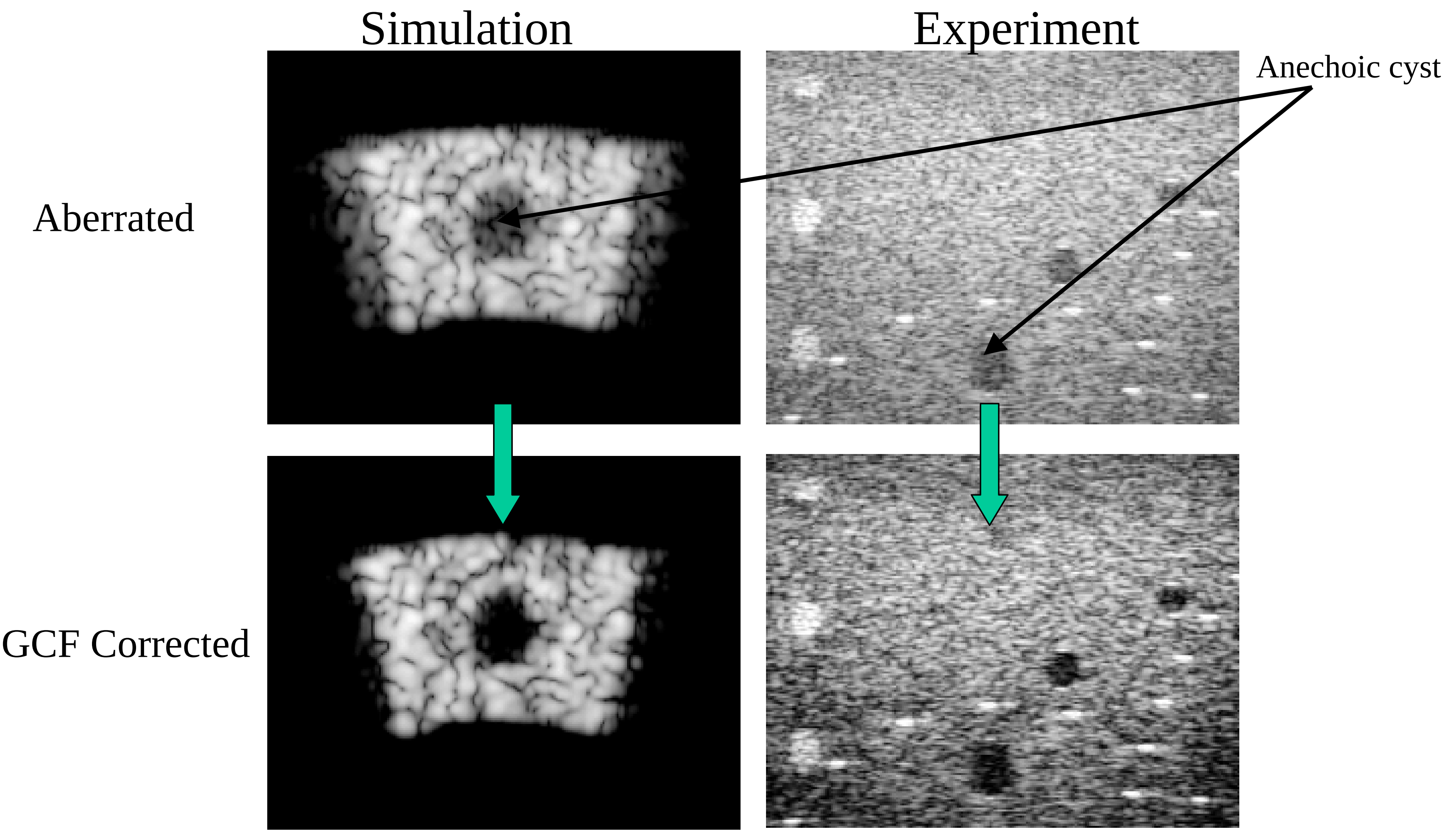
No aberration



- Lower GCF should be used to reduce the image data because significant beamforming errors are present

Aberrated

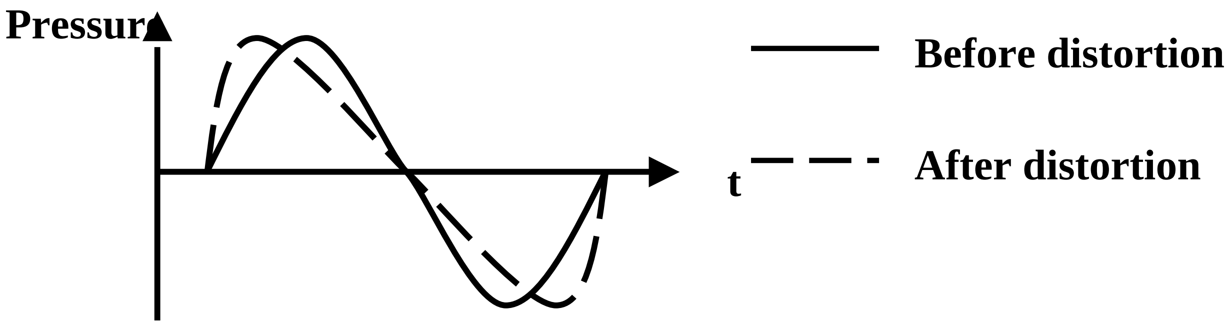
Results


Simulation
Experiment
Anechoic cyst
Aberrated
GCF Corrected

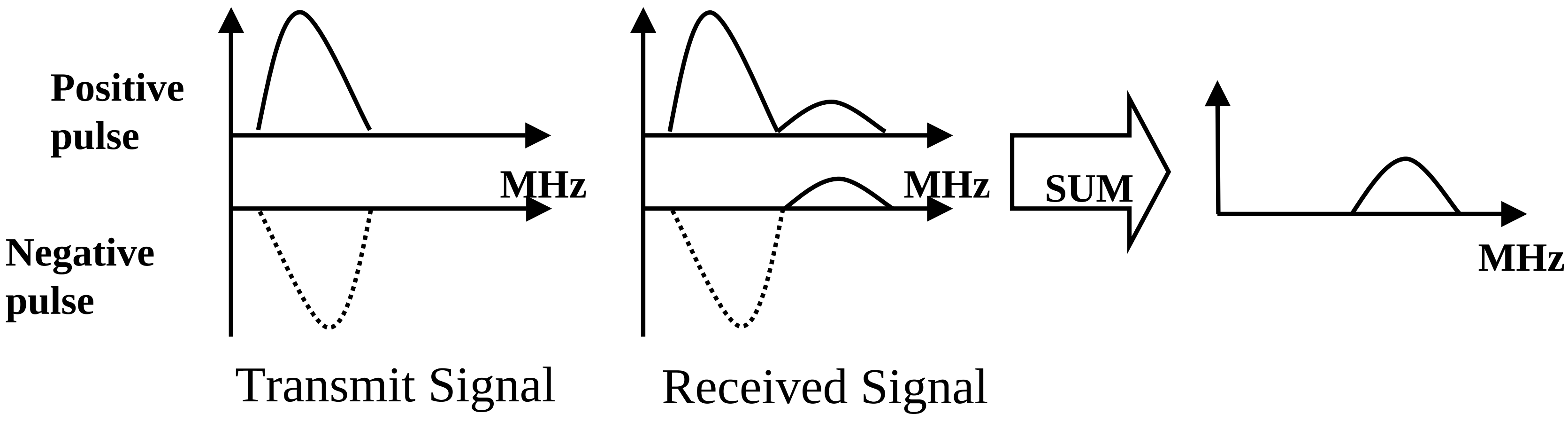
2. Ultrasonic Nonlinear Imaging

Ultrasonic Tissue Harmonic Imaging

- **Tissue Harmonic**
  - the harmonic component generated from *finite amplitude distortion*

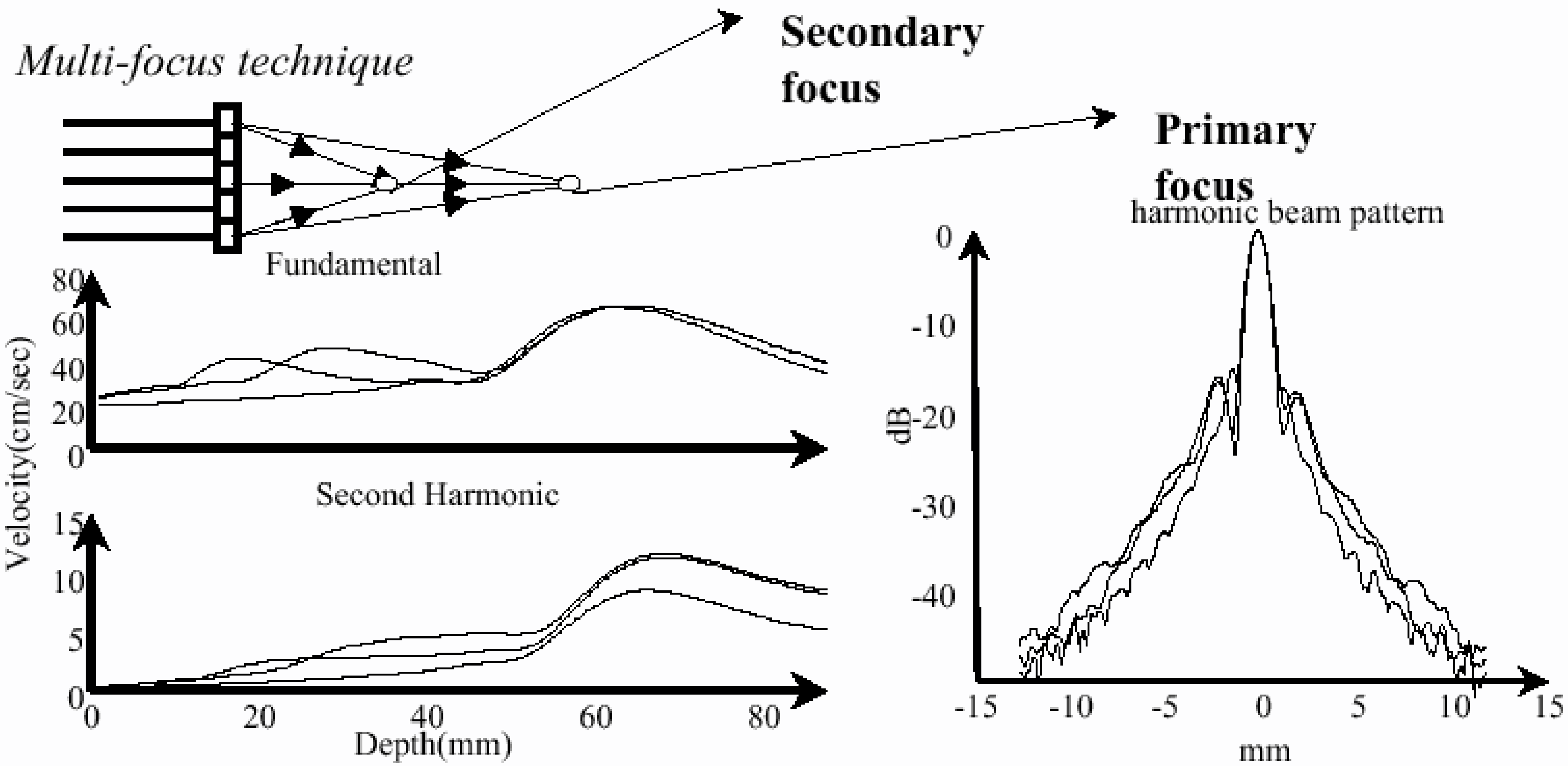


- **Pulse Inversion**
  - Better fundamental rejection, lower frame rate



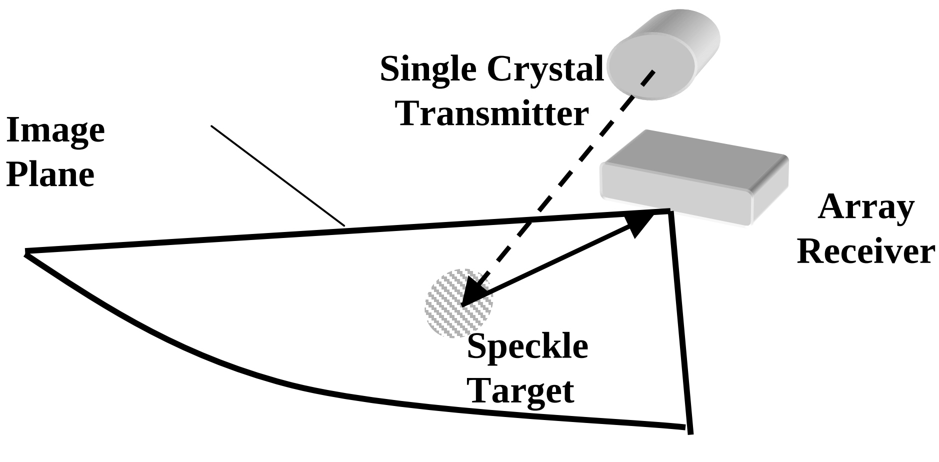
Ultrasonic Tissue Harmonic Imaging

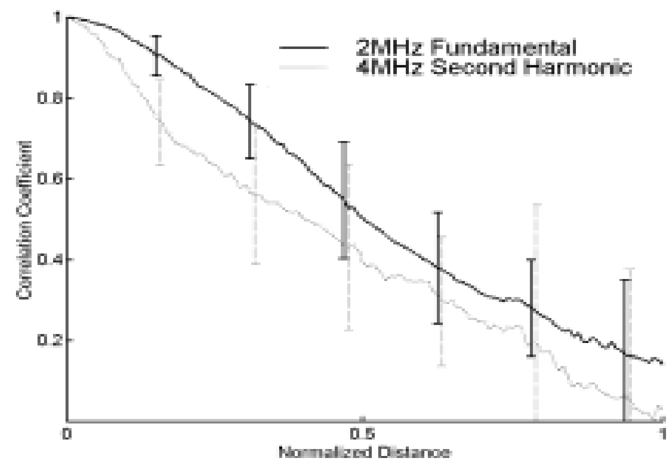
- The Effect of Multi-focus Technique on Tissue Harmonic Image
- Effects of Harmonic Leakage on Tissue Harmonic Imaging

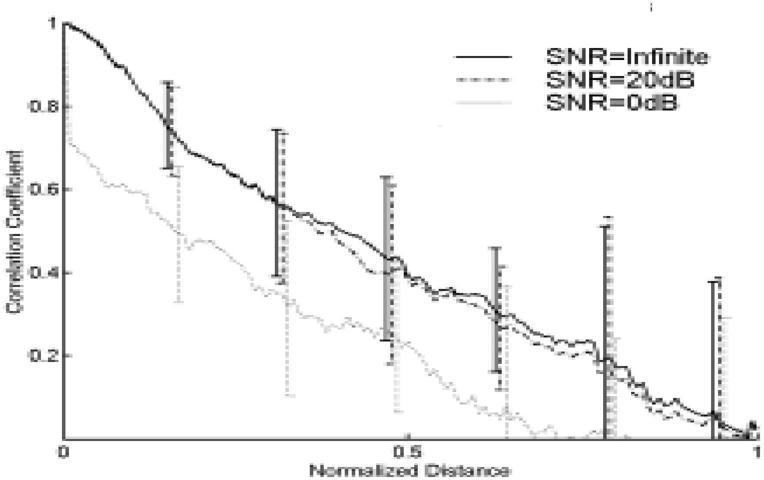


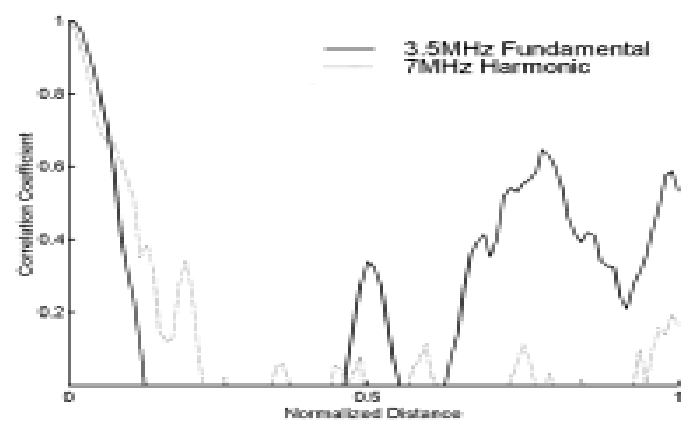
Ultrasonic Tissue Harmonic Imaging

- **Harmonic spatial covariance analysis**
  - Effects of SNR
  - Effects of sound velocity inhomogeneities



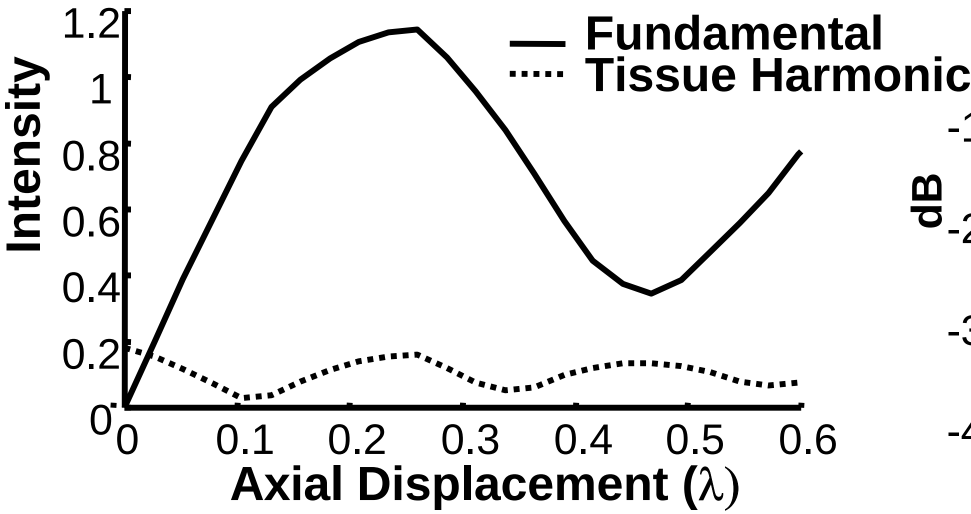


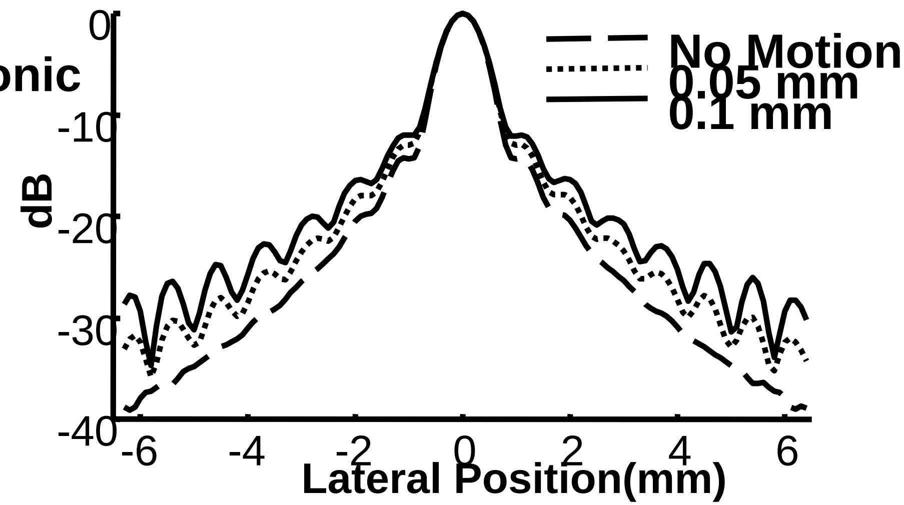


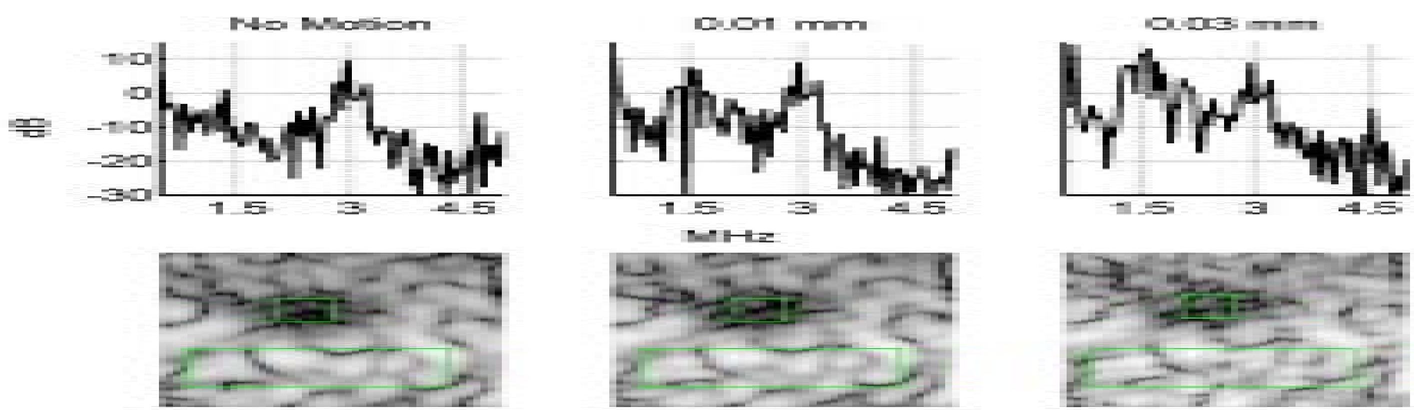


Ultrasonic Tissue Harmonic Imaging

- **Motion artifacts of Pulse Inversion Technique**
  - Effects of SNR
  - Effects of sound velocity inhomogeneities







3. Ultrasonic Elastic Imaging

Ultrasonic elastic imaging

- Ultrasonic strain compounding image based on a fast speckle tracking algorithm
- Strain compounding technique
  - Improve contrast resolution of the image
  - Steps:

    Obtain an uncompressed image as a basis

    Applying an external force on the object yields deformation

    Modify the deformation in the image plane

    Average the modified and original images

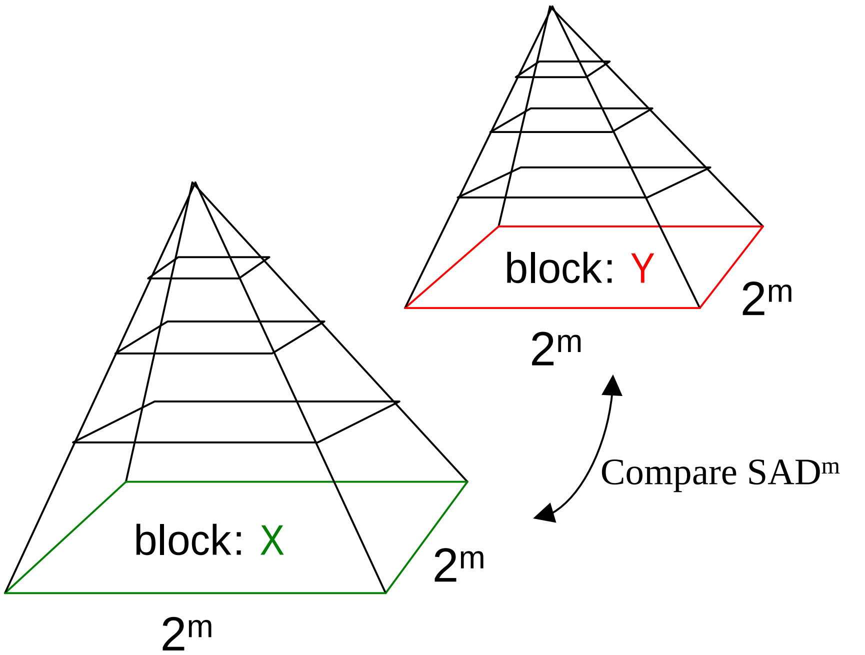
2D fast speckle tracking algorithm

- Block Sum Pyramid
  - Take threshold: SADmin
  - Reduce the computations of *SAD*

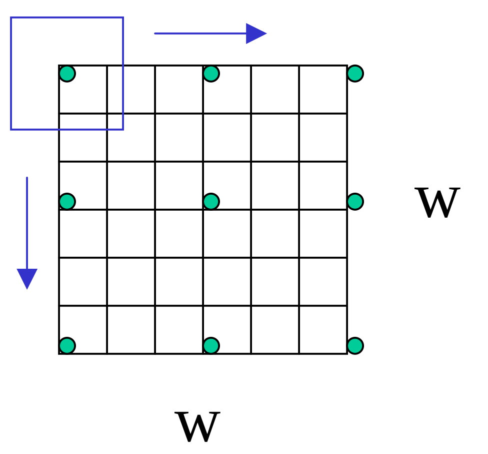
Pyramid structure

$$X^{m-1}(i,j) = X^{m}(2i-1,2j-1) + X^{m}(2i-1,2j) + X^{m}(2i,2j-1) + X^{m}(2i,2j)$$

$$SAD^{m}(X,Y) = \sum_{i=1}^{2^m}\sum_{j=1}^{2^m}\left|X^{m}(i,j) - Y^{m}(i,j)\right|$$



- Multilevel Block matching
  - Reduce *numbers* of points to be searched
  - We use 2 levels
  - 1^{st} level: window size= w*w, 9 points
  - 2^{nd} level: window size= ½*w* ½ *w, all points



1st level

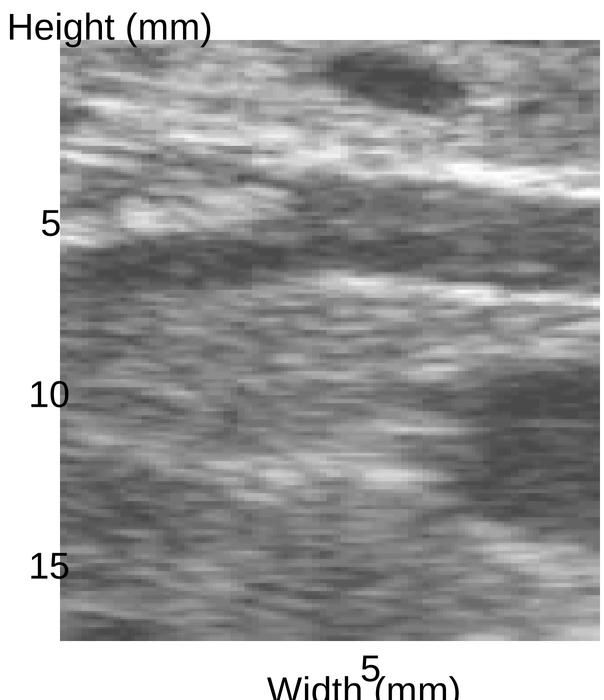
Results

- Algorithm performance For 121 pixels

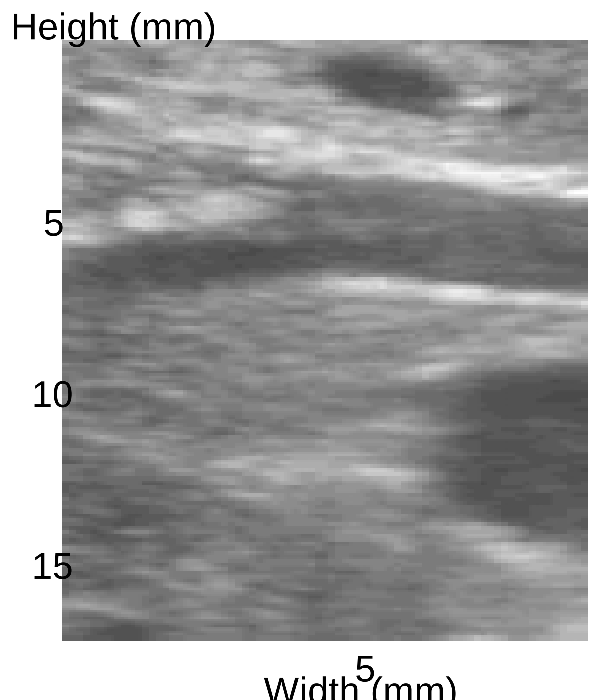
Language	FSA	BSPA	BSPA & Multilevel	Ratio
Matlab	27.14 (s)	14.17 (s)	7.64 (s)	3.6 : 2 : 1
C	12.3 (s)	1.53 (s)	0.998 (s)	12 : 1.5 : 1

BSP & Multilevel algorithm is indeed not only faster than traditional algorithm, but also as accurate as traditional one.

- Compounding image



L: original image



R: compounding image with BSP & Multilevel

Liver

SNR=0.1164

SNR ↑

Computing speed ↑

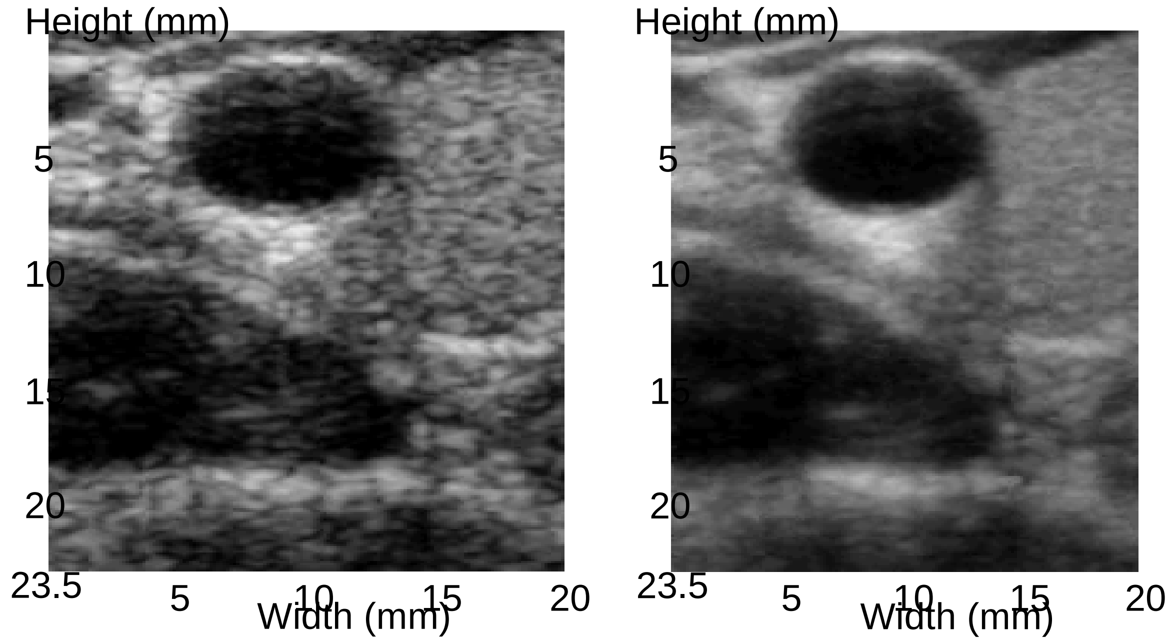
Cont’d

Thyroid

SNR=0.9625

SNR ↑

Computing speed ↑



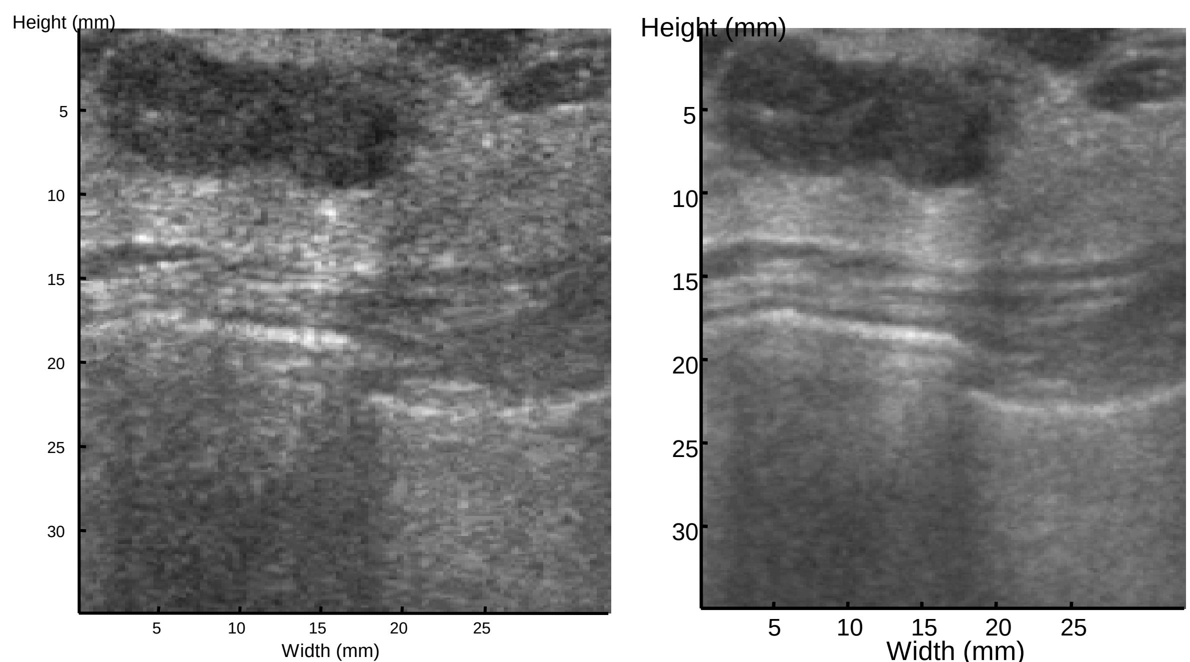
L: original image R: compounded image with BSP & Multilevel

Breast

SNR=0.1692

SNR ↑

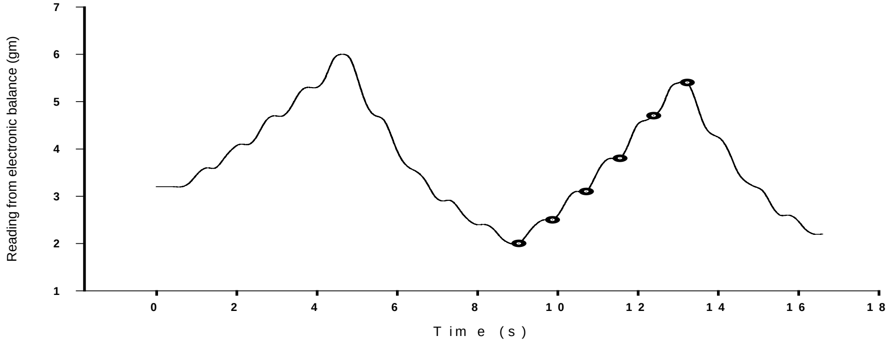
Computing speed ↑



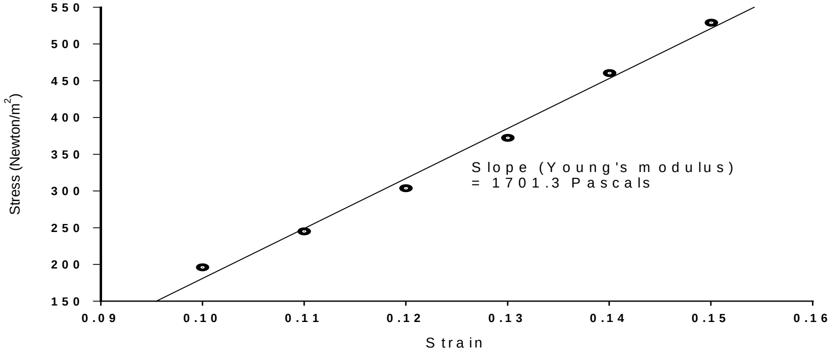
Young's Modulus Measurements of Human Liver and Correlation with Pathological Findings

The Experimental Set-up

The Readings from Electrical Balance as a Function of Time



The Stress- Strain Curve

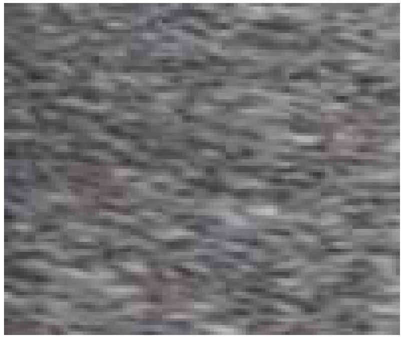


The Young's modulus of normal liver, cirrhotic liver and hepatic tumors

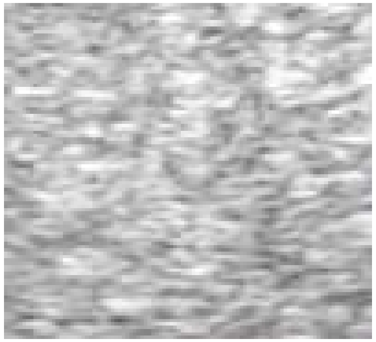
Preload strain	Young's Modulus (Pascals)						
	Liver parenchyma (mean value)			Hepatic tumors			
	Normal	Cirrhosis (Fibrosis score: 4)	Cirrhosis (Fibrosis score: 5)	Hepatocellular carcinoma	Cholangio-carcinoma	Focal nodular hyperplasia	Hemangioma (Emelianov et al. 1998)
5%	642.6	1106.4	1649.0	Smaller than normal liver	3003.7	1084.9	Larger than normal liver
10%	1083.6	2373.6	4930.7		12098.1	2522.5	

Tissue characterization of Ultrasonic B-image

- Compare normal with cirrhotic liver
  - Statistical method
  - Conventional and non-separable wavelet decomposition method



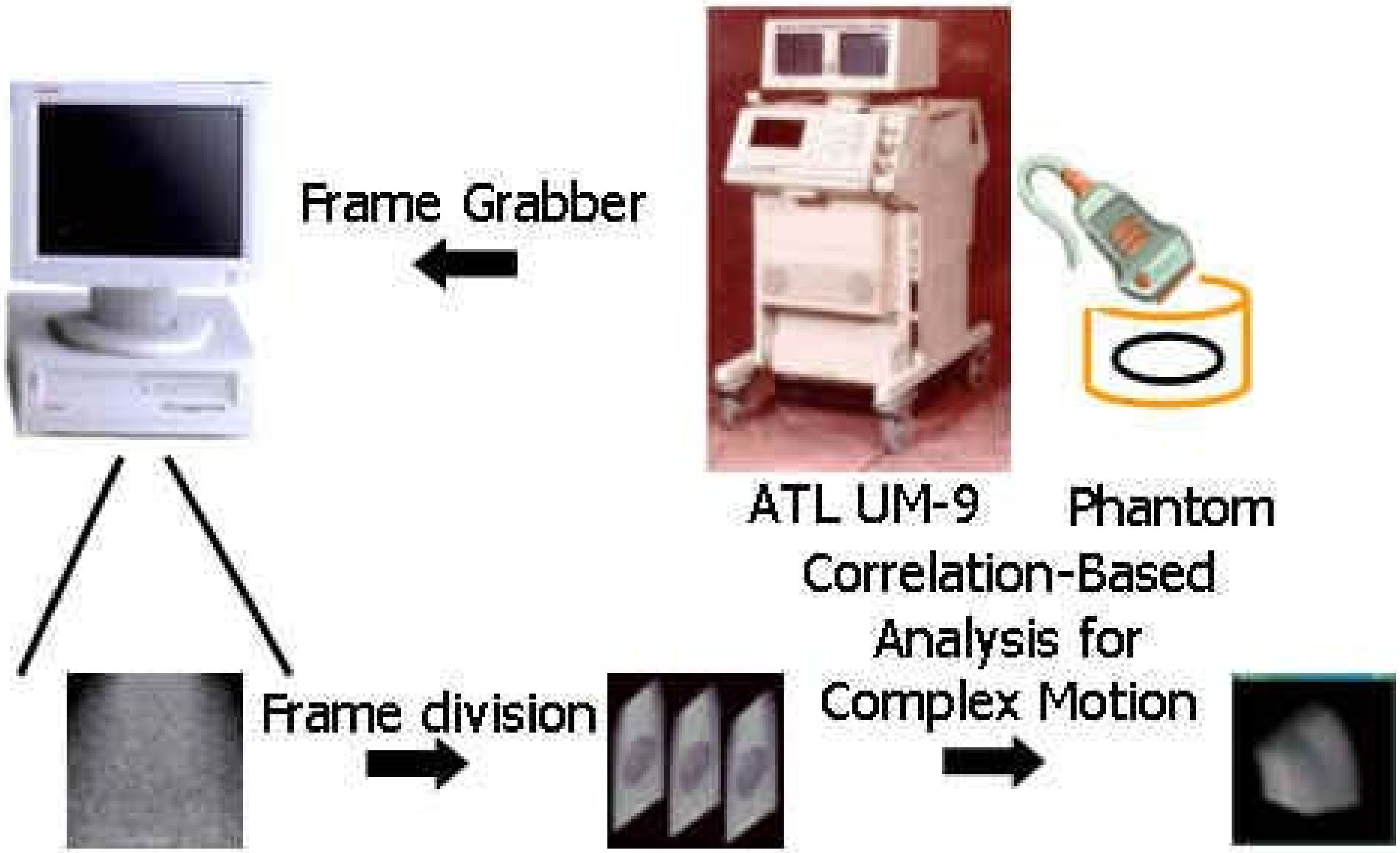
Normal liver



Cirrhotic liver

4. 3-D Ultrasound Imaging

A Free-Hand 3D Ultrasound Imaging System



Correlation-Based Analysis for Complex Motion

Acquire 2 images

⇩ Speckle tracking

Find x'、 z' and α'

⇩ Correct the x' and z' motions

Images are spatial matched

⇩

Calculate 2D C.C.

⇩ Go through all β, γ, Y combinations

Find the best matched combination

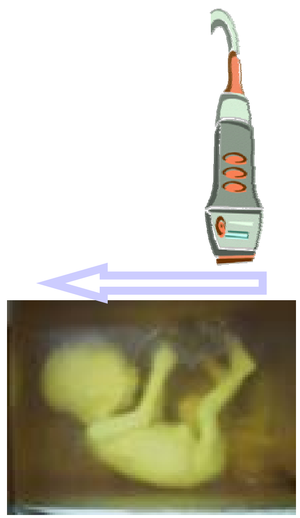
⇩

x'、 y'、 z'、 α'、 β'、 γ'

A 3D System Integration

- Platform：

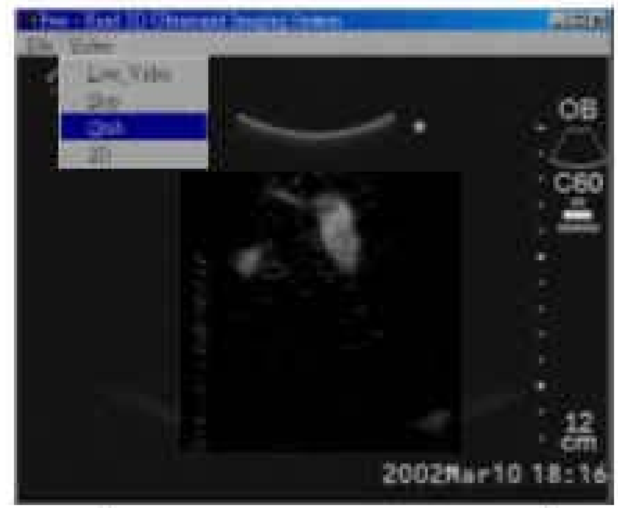
- Win NT、OpenGL



Baby Phantom

ATL UM-9

Frame Grabber

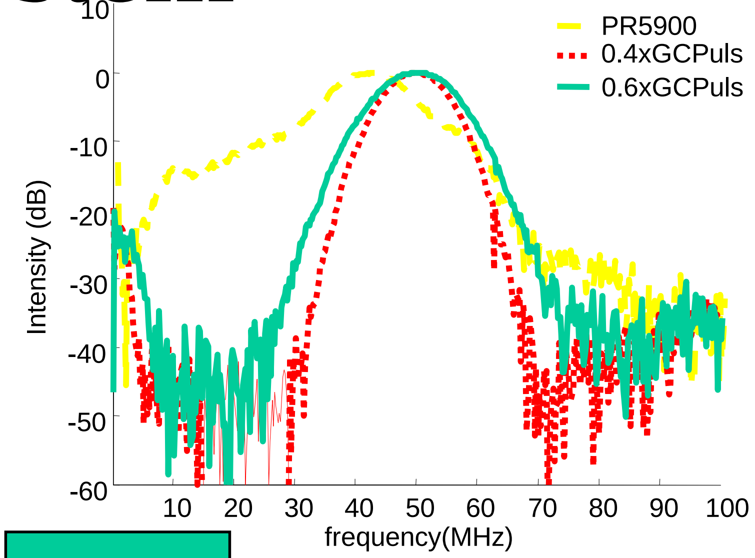


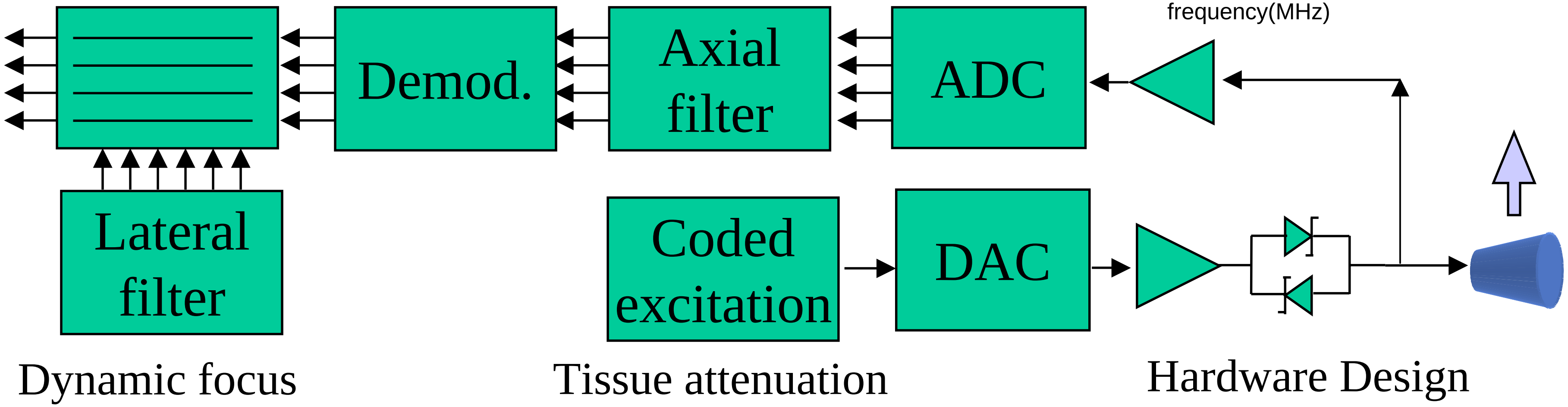
3D Rendering

5. High Frequency Ultrasonic Imaging

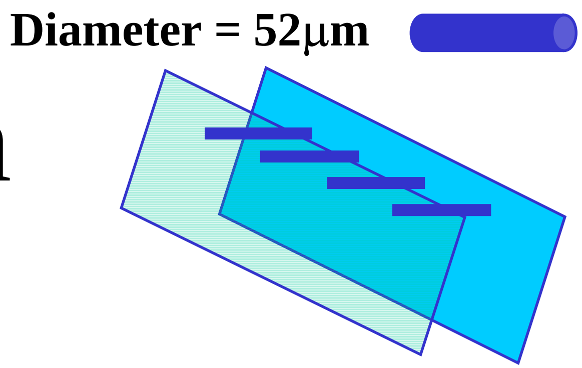
High frequency ultrasonic imaging system

- Fully digital system architecture
- 50 MHz center frequency
- 60 % fractional bandwidth

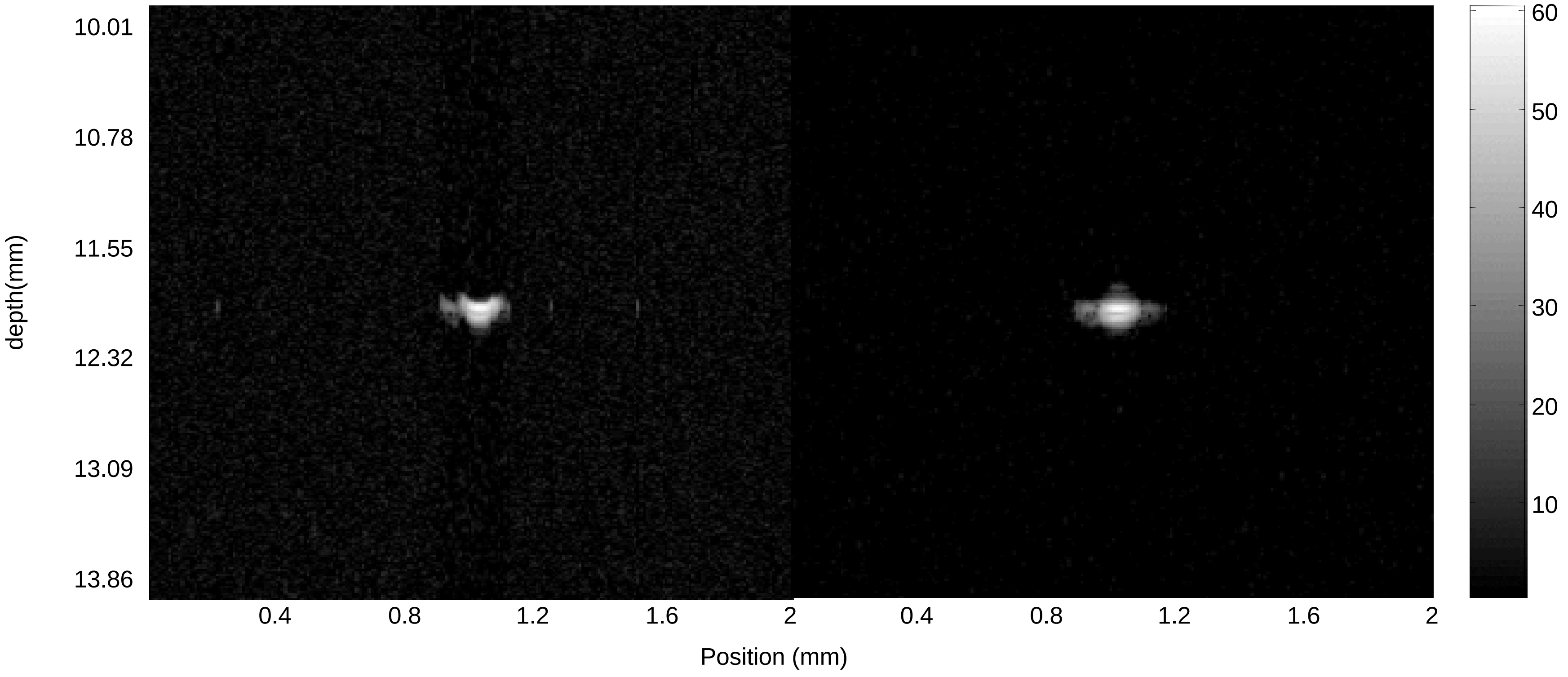




Wire phantom

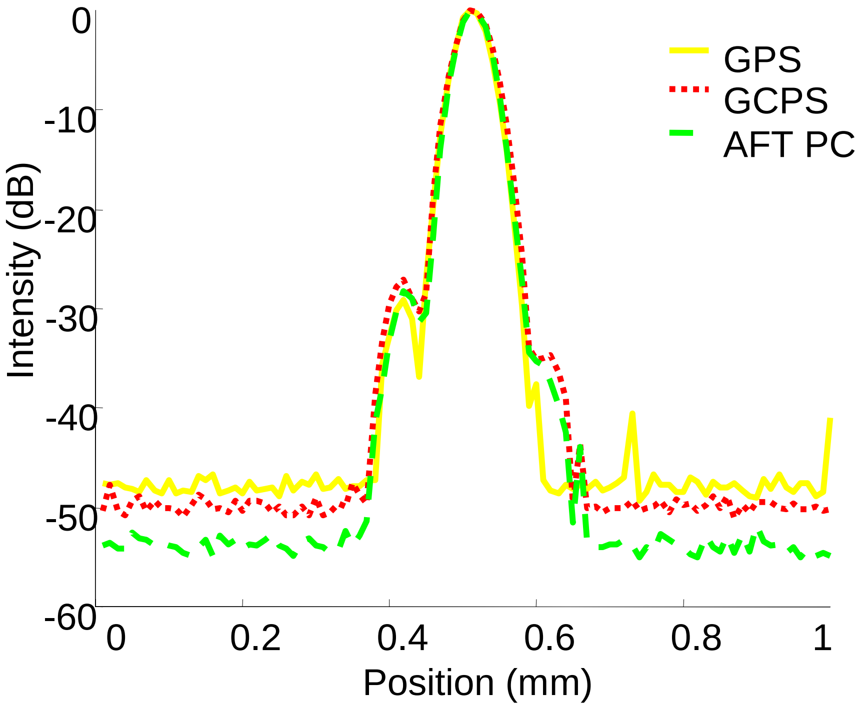


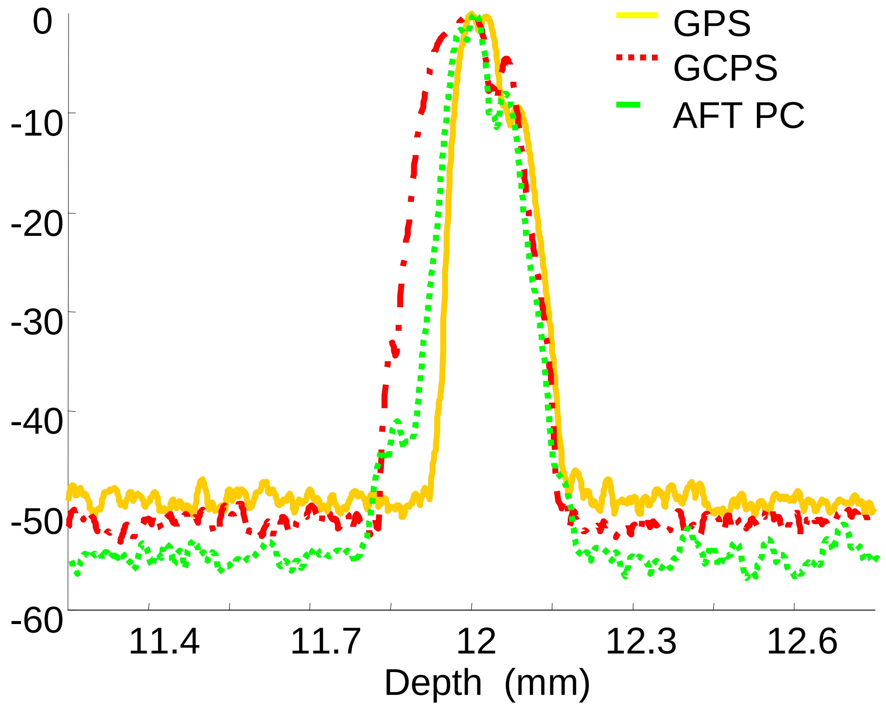
- 52um nylon wire phantom
- Gaussian Pulse Gaussian Chirp Pulse



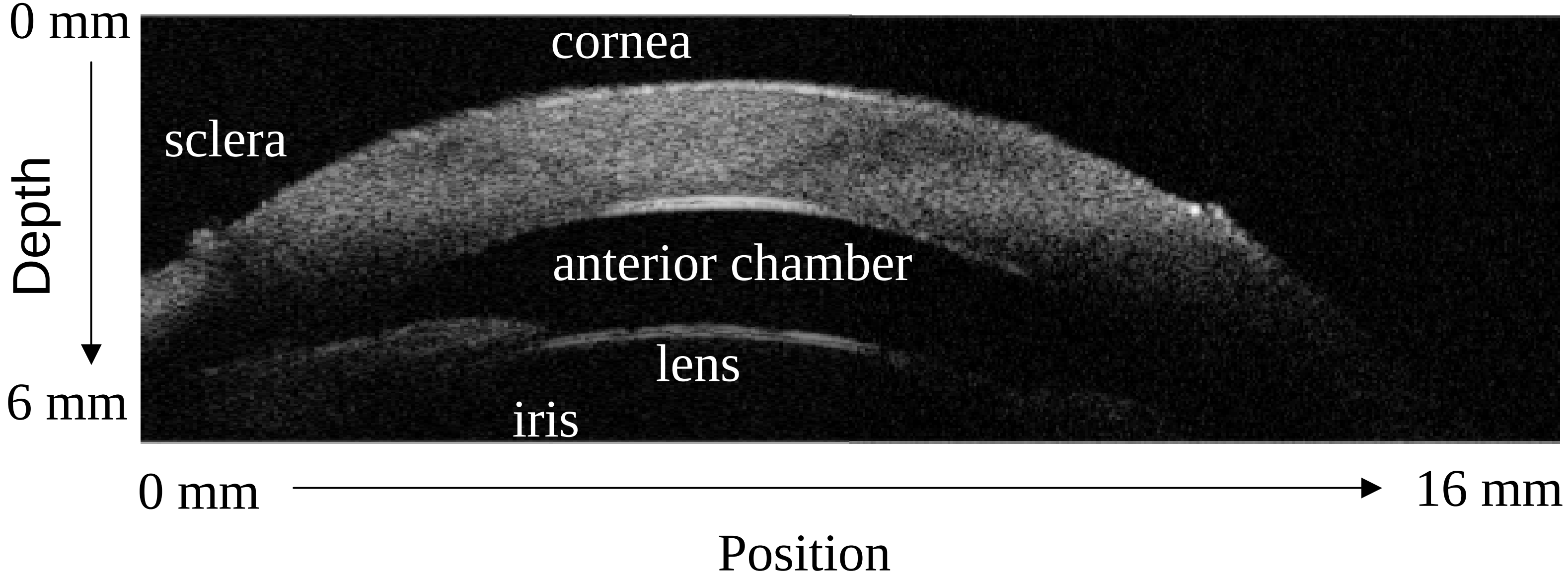
Resolution test

- Lateral projection: axial projection
- Spatial resolution is about 60 μm



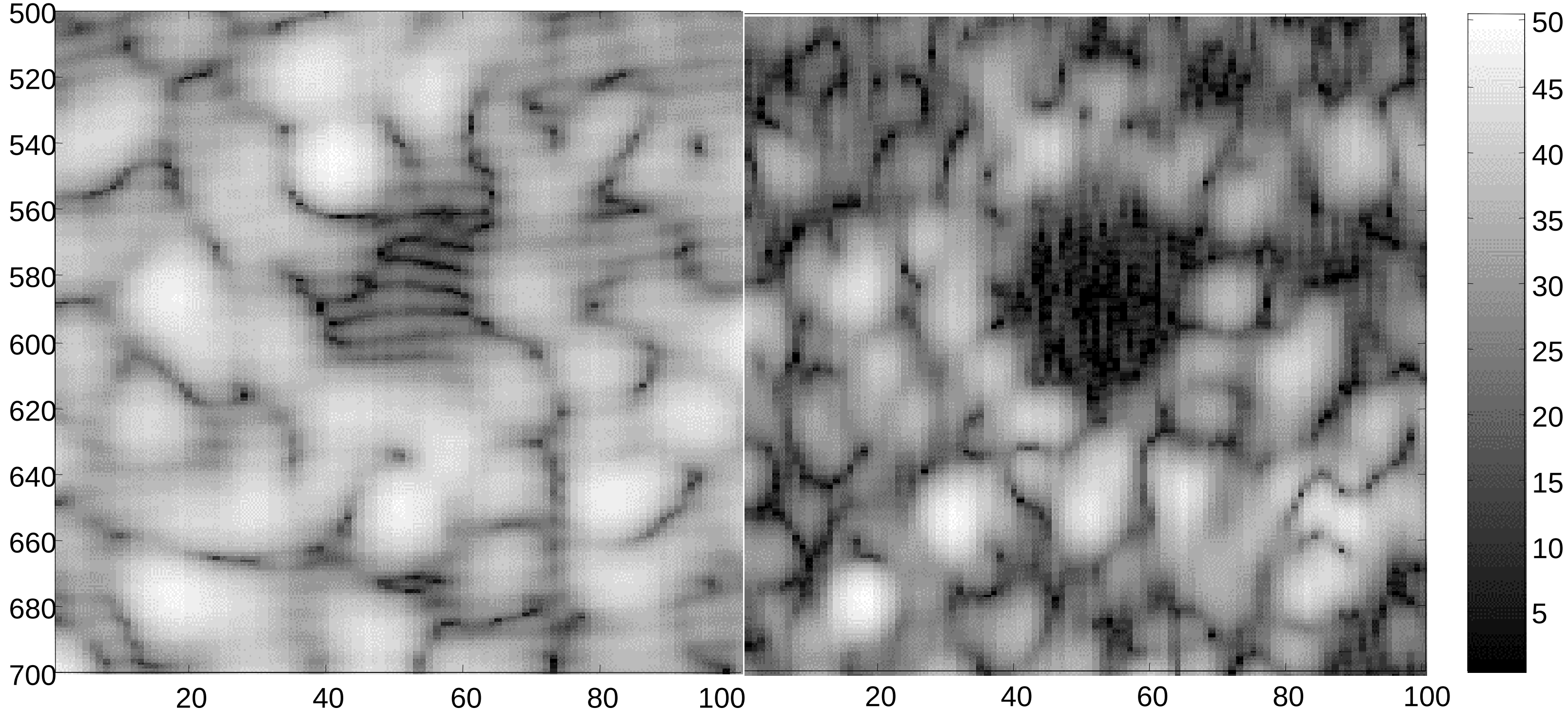


in-vitro pig eye image



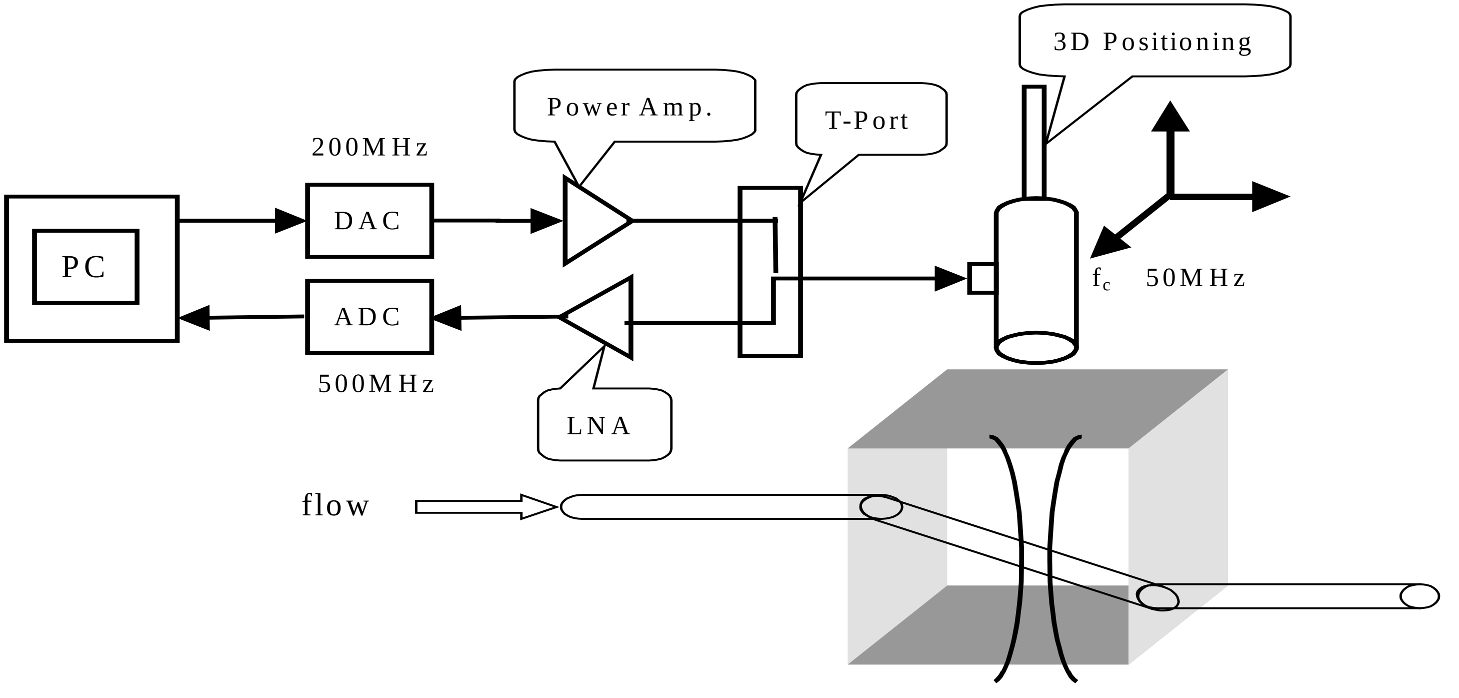
Tissue harmonic imaging

- Pulse inversion technique cancels fundamental signal



High Frequency Ultrasound Doppler

- 50MHz High Frequency Ultrasound : wideband transmitted signal (short transmitted pulse) and narrow lateral beamwidth → better spatial and velocity resolution (down to mm/s), capable of estimating low velocities blood flow in small vessels.

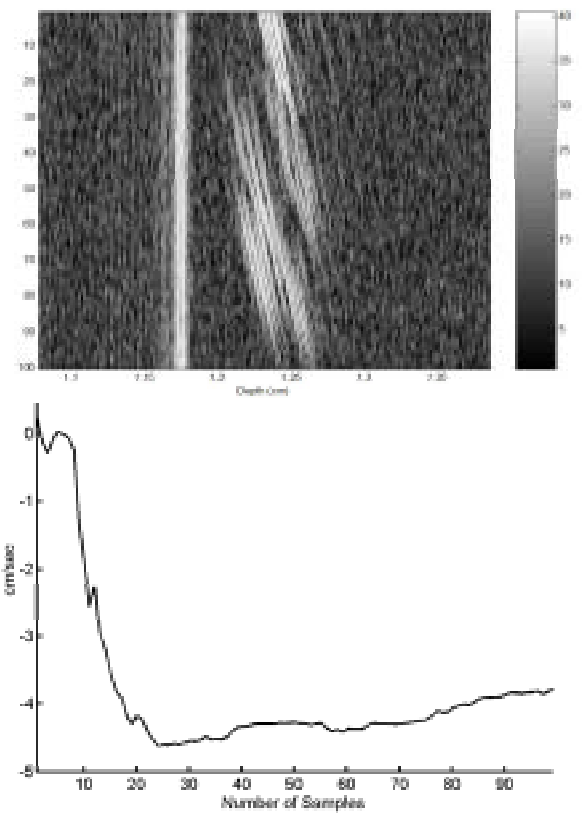


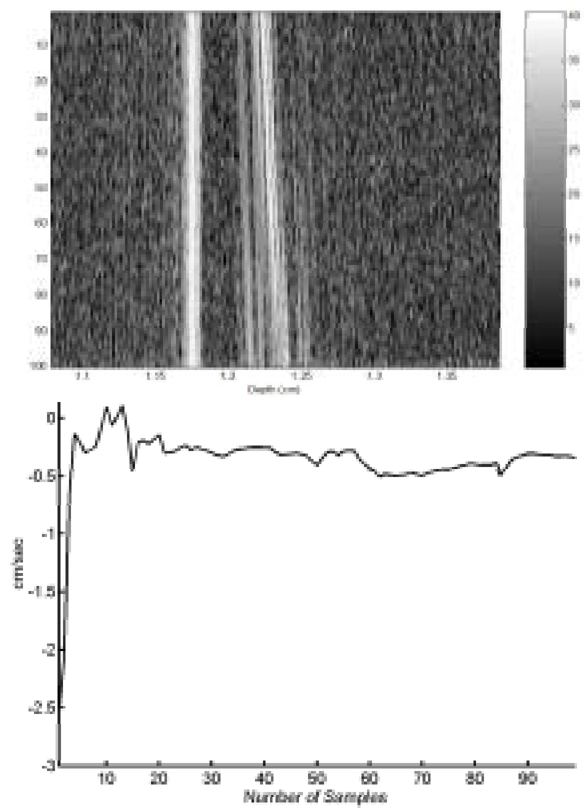
High Frequency Ultrasound Doppler

In-Vitro Flow Estimation : Autocorrelation Technique

$$R(t) \equiv \int_{-\infty}^{\infty} S(t+\tau)\, S^*(\tau)\, d\tau$$

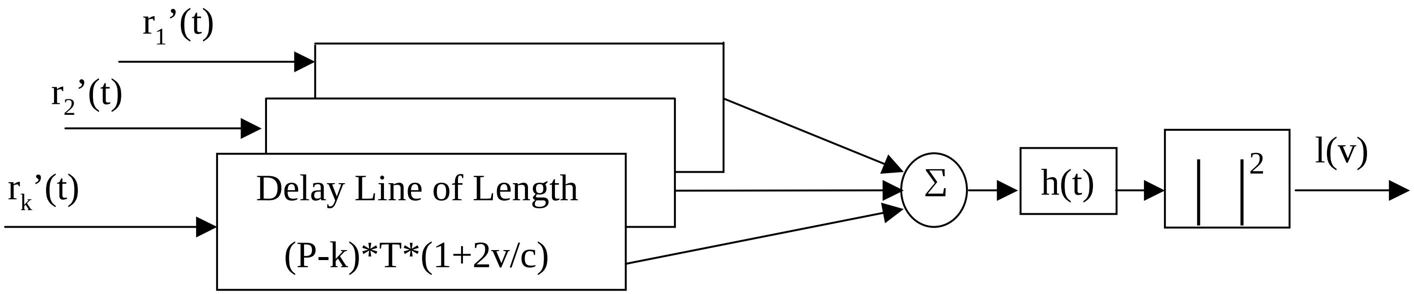
$$R(t) = \left|R(t)\right| e^{j\theta(t)} \longrightarrow \bar{f} = \frac{\theta(T)}{2\pi T}$$, T : PRI



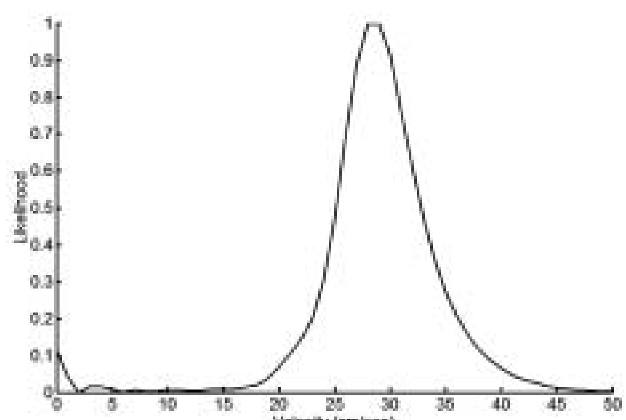


High Frequency Ultrasound Doppler

In-Vitro Flow Estimation : WMLE Technique

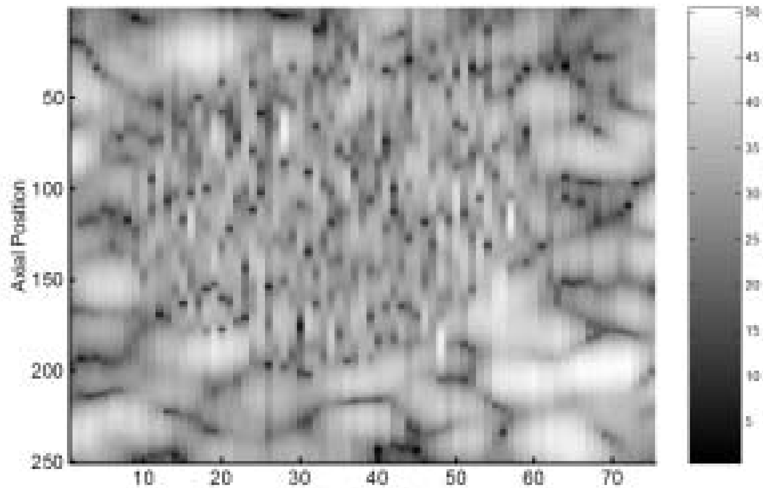


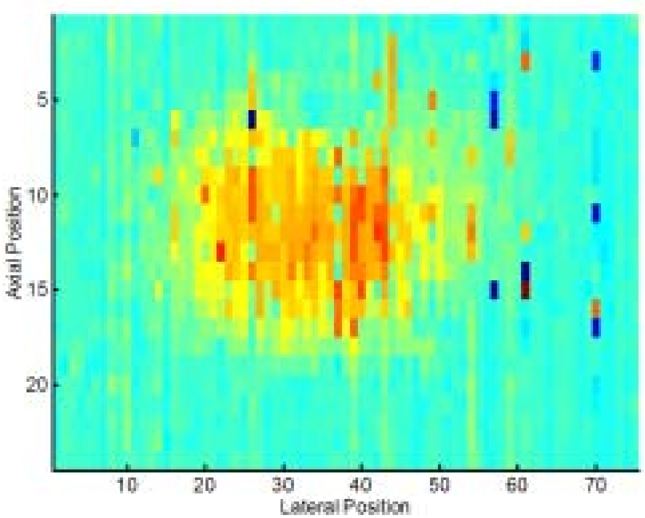
Bank of delay lines, and filter h(t) matched to the expected demodulated echo signals which correspond to various velocities. The maximum likelihood velocity is then given by the filter with the largest output.

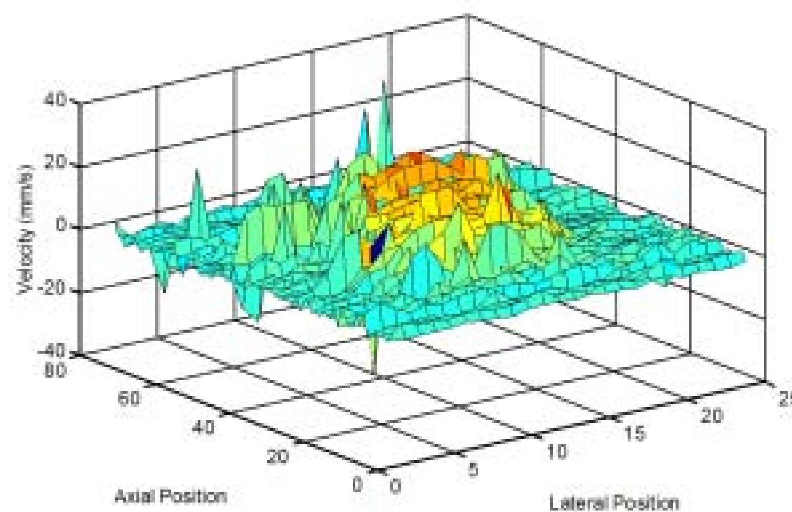


High Frequency Ultrasound Doppler

In-Vitro 2D Flow Data : 500μm diameter cyst,
with maximum velocity   20mm/s

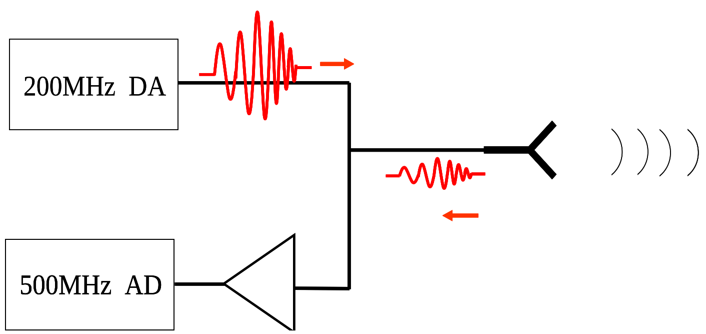




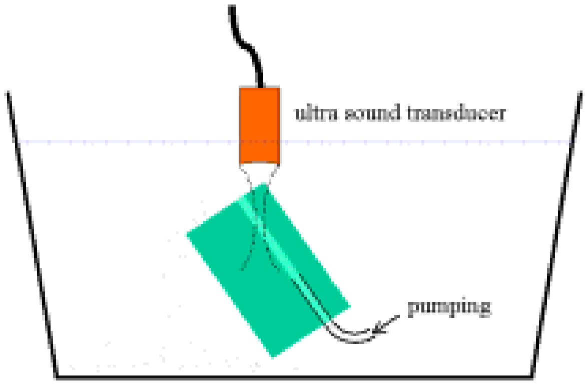


High Frequency Ultrasound Experiment
25~50MHz

Basic System Diagram



Flow System



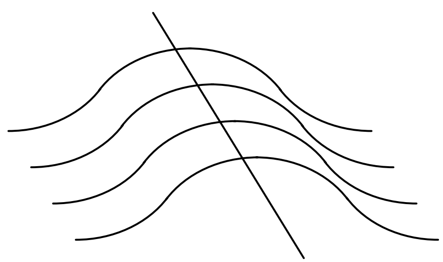
Experiment Condition

200MHz DA excite 25~50 MHz coded ultrasound wave
500MHz AD receiving
Wide band ultrasound transducer
Up to 20KHz PRF

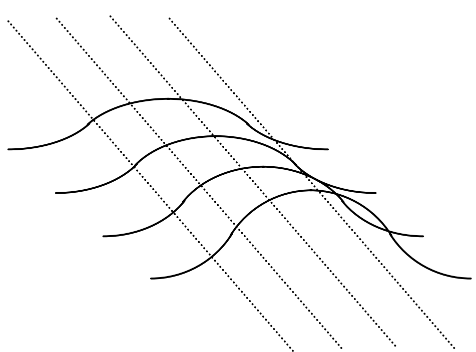
High Frequency Flow Estimation

RF Butterfly Search (Multi-line)

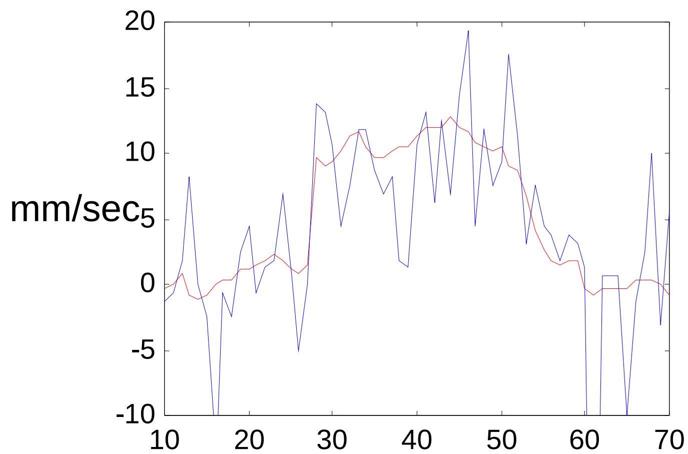
Traditional Butterfly Search Line



Multiple Butterfly Search Lines



Experimental λ-Flow Result

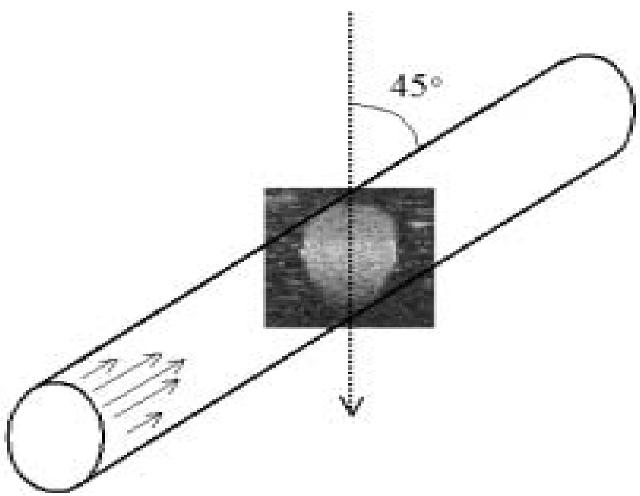


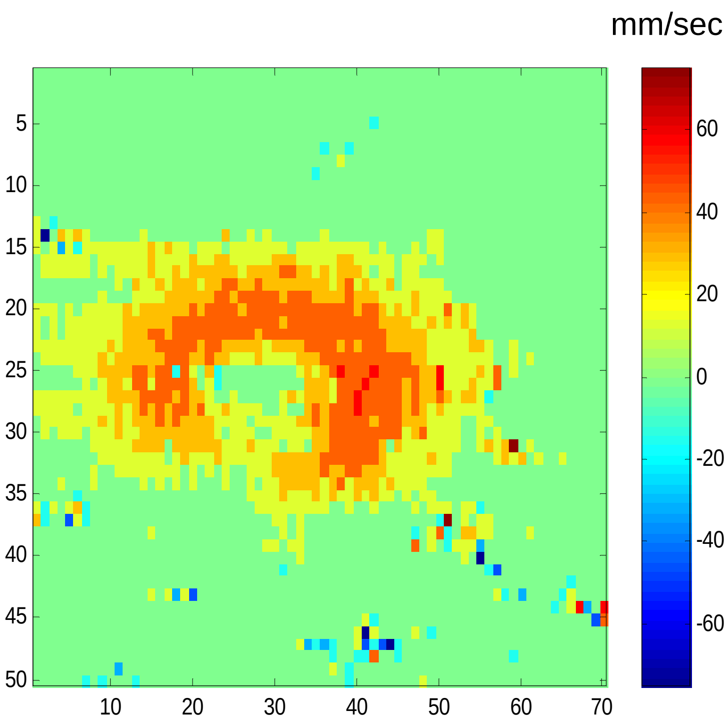
Traditional RF Butterfly

Multi-line Butterfly

High Frequency Flow Estimation

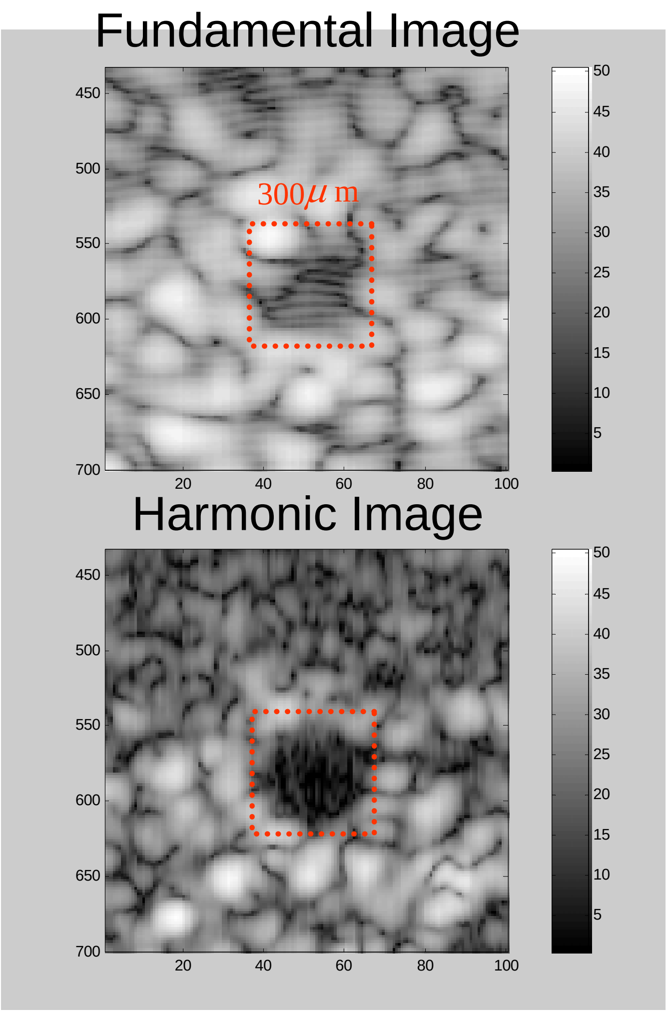
Color Flow Image

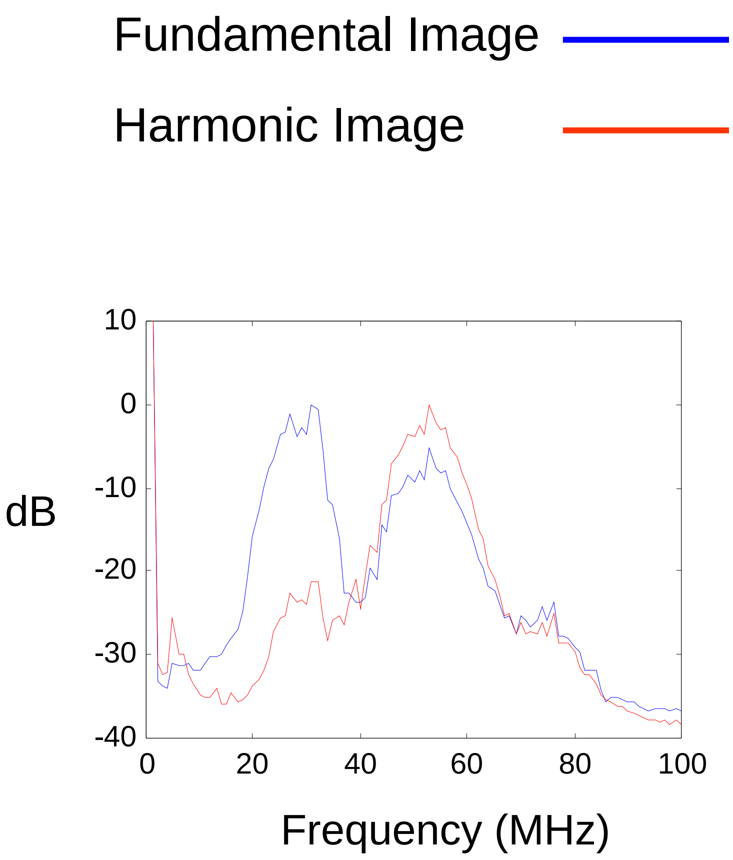




High Frequency Harmonic Image

25MHz 300 μm Cyst image

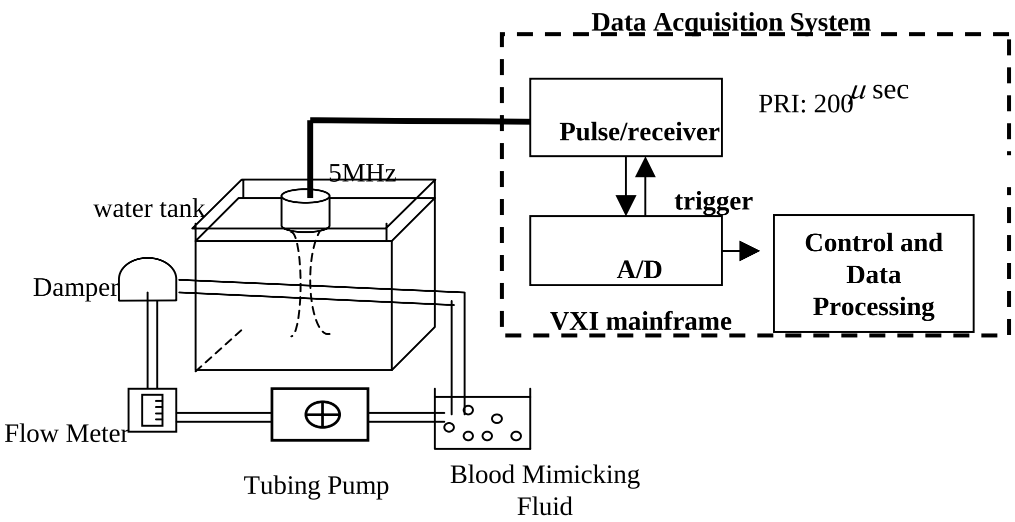


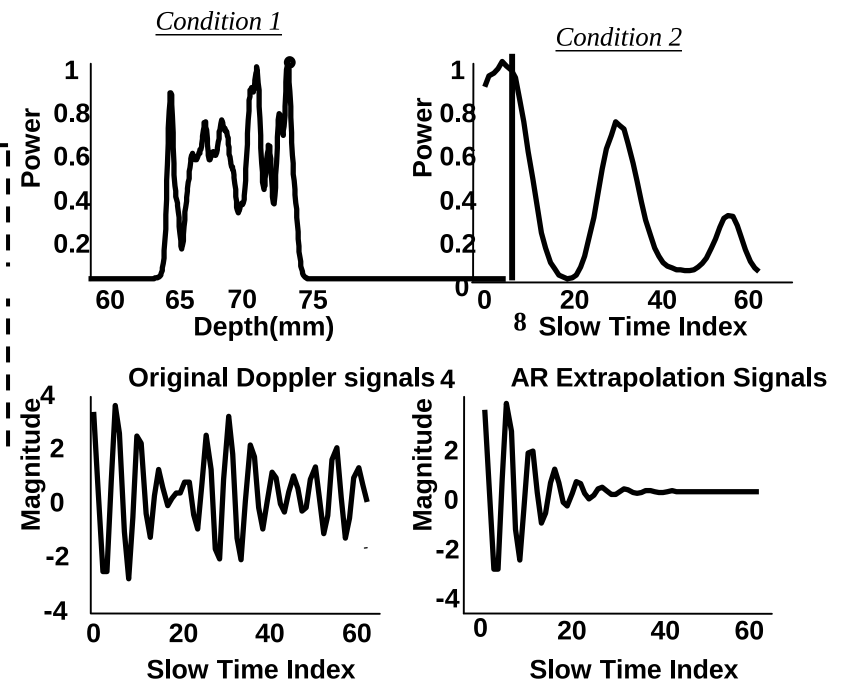


6. Blood Flow Estimation Using Ultrasonic Contrast Agent

Doppler Blood Flow Estimation (I)

- Doppler Angle Estimation
- Blood Flow Estimation

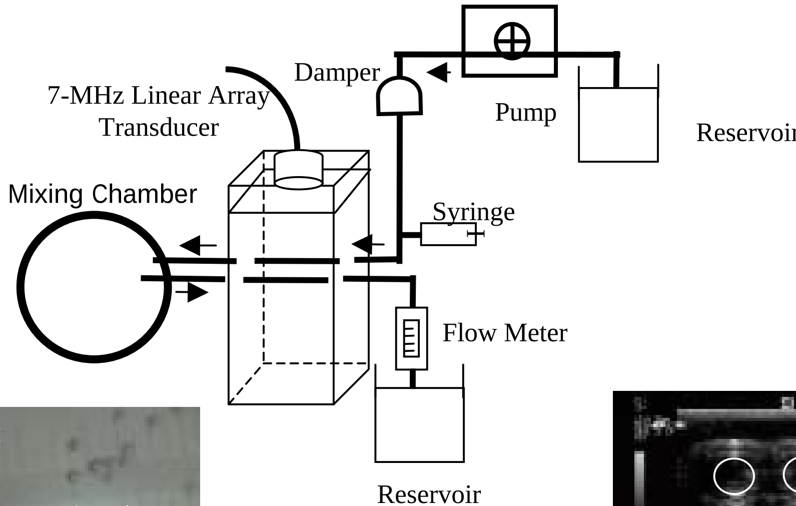


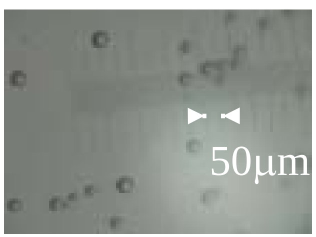


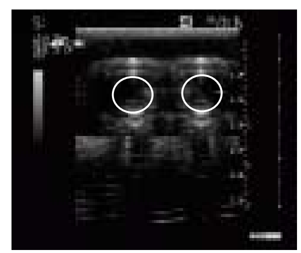
1. P.-C. Li, C.-J. Cheng and C.-C. Shen, “Doppler Angle Estimation Using Correlation”, IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control, Vol. 47, No. 1, pp. 188-196, 2000.
2. P.-C. Li, C.-J. Cheng and C.-K Yeh, "On the Velocity Estimation Using Speckle Decorrelation", IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control, Vol. 48, No. 4, pp. 1084-1091, July, 2001.
3. C.-K. Yeh and P.-C. Li, "Doppler Angle Estimation Using AR Modeling", IEEE Transactions on Ultrasonics, Ferroelectrics and Frequency Control. June, 2002

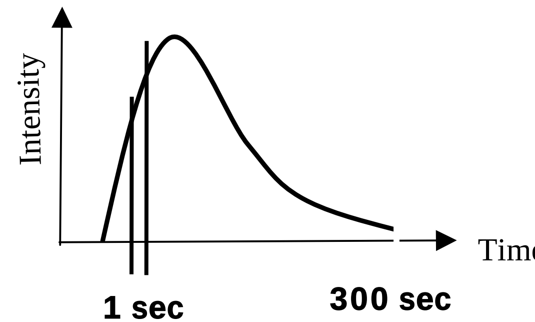
Blood Flow Estimation Using Ultrasonic Contrast Agent

- Blood Flow Estimation (Indicator-Dilution Theory)
- Time-Vary Method









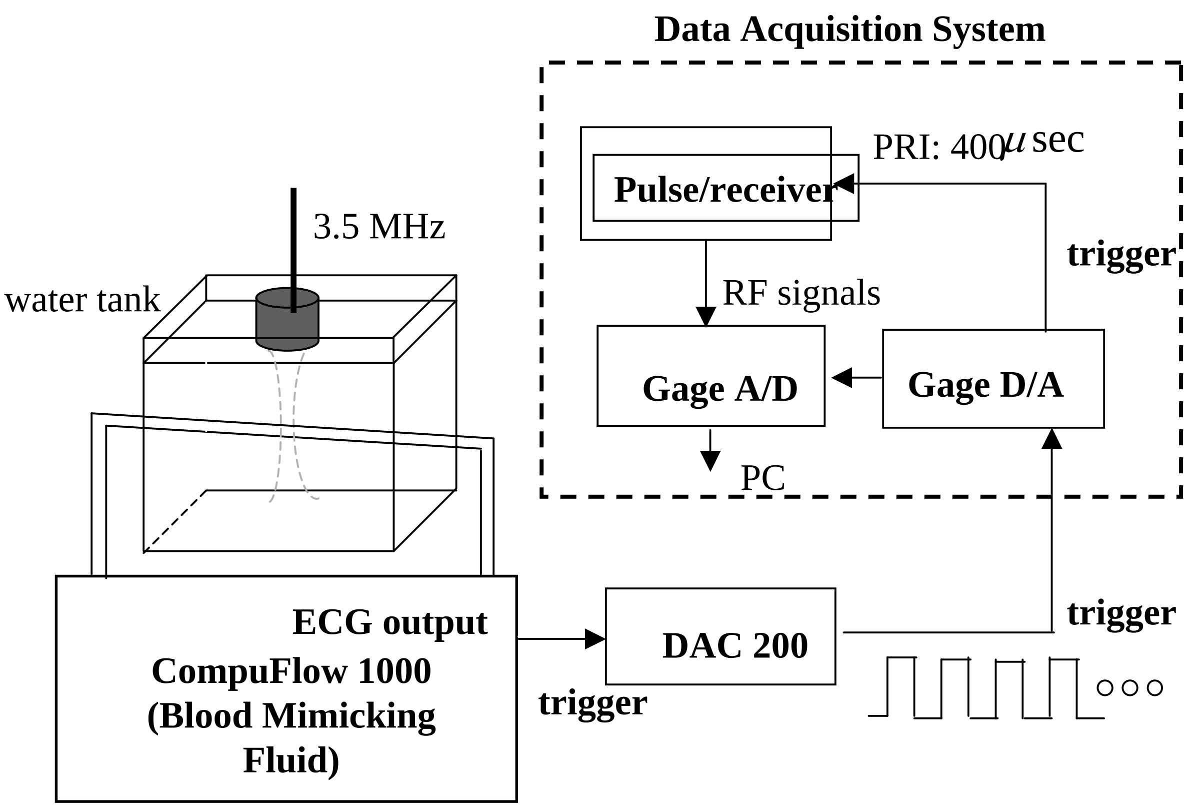
Time-Intensity Curve (TIC)

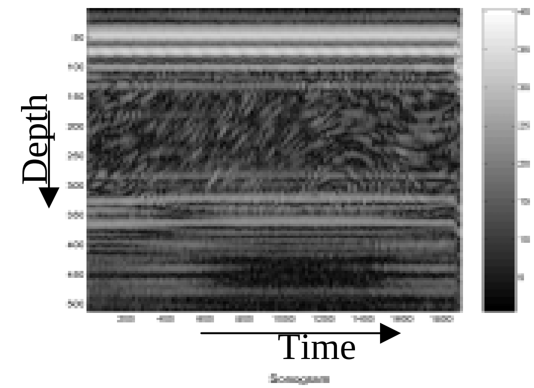
1. C.-K. Yeh S.-W. Wang and P.-C. Li, Feasibility Study on the Time-Intensity Based Blood Flow Measurements Using Deconvolution," *Ultrasonic Imaging*, vol. 23, pp. 90-105, April, 2001,

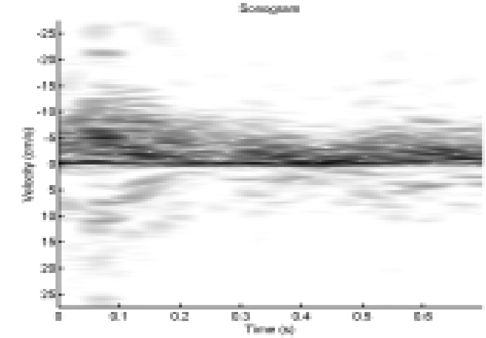
2. P.-C. Li, C.-K. Yeh and S.-W. Wang, "Time-Intensity Based Volumetric Flow Measurements: An In Vitro Study", Ultrasound in Medicine and Biology. vol. 28, no. 3, pp. 349–358, 2002

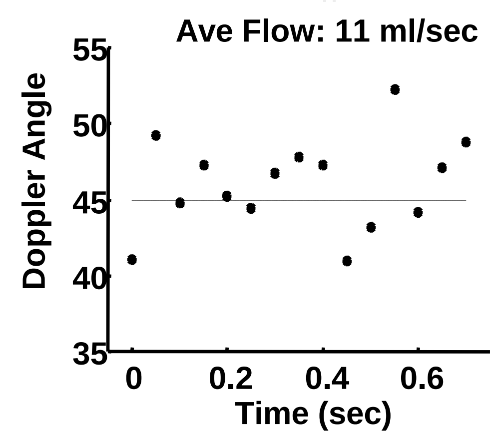
Doppler Blood Flow Estimation in Pulsatile Flow (II)

- Doppler Angle Estimation
- Blood Flow Estimation





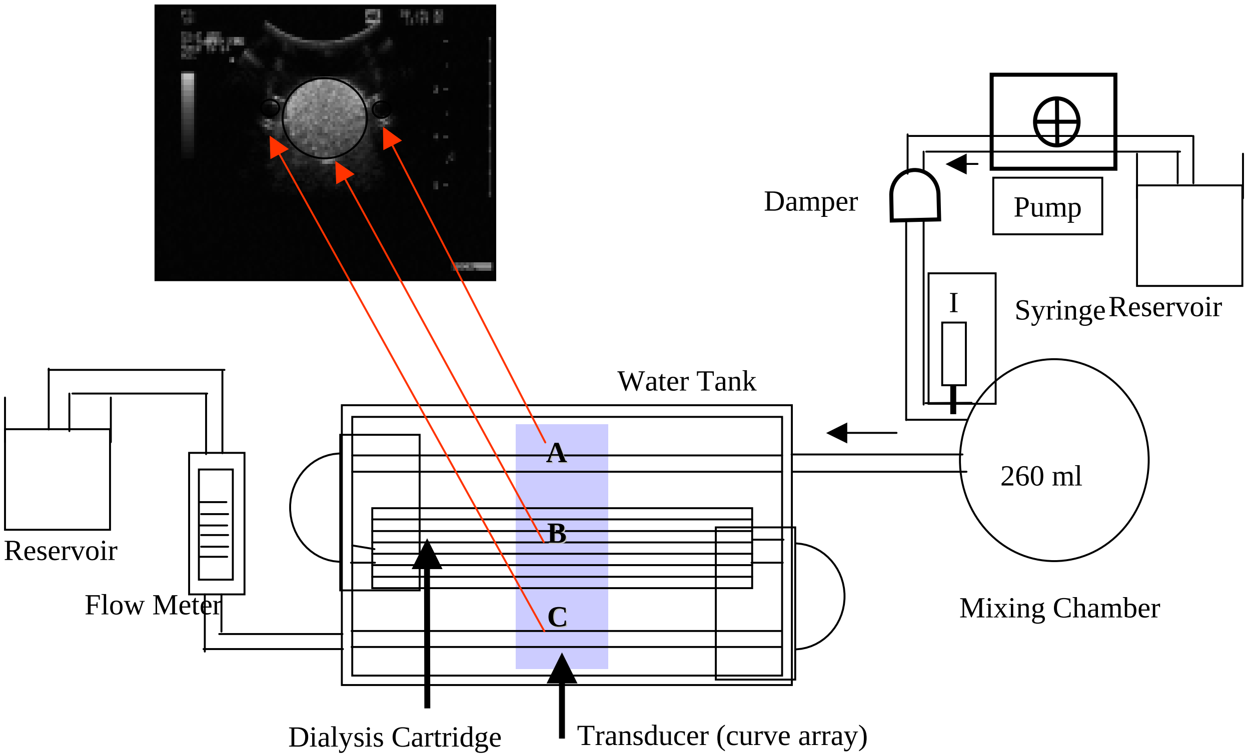


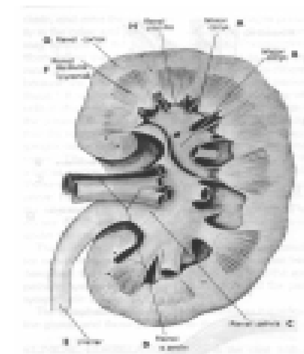


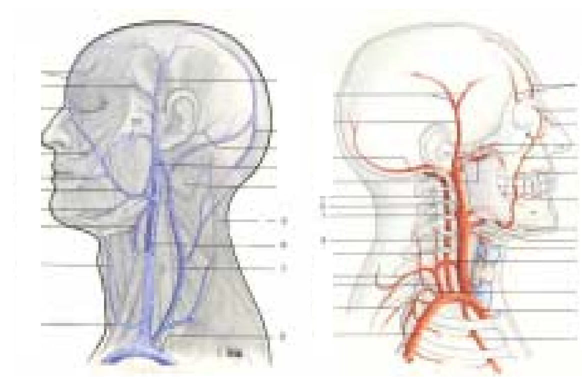
Chih-Kuang Yeh and P. C. Li, "Doppler Angle Estimation of Pulsatile Flow Using AR Modeling," *Ultrasonic Imaging*, 2002

Blood Flow Estimation Using Ultrasonic Contrast Agent

- Shadowing Effect
- Input and Output Time-Intensity Curves (IOTIC)



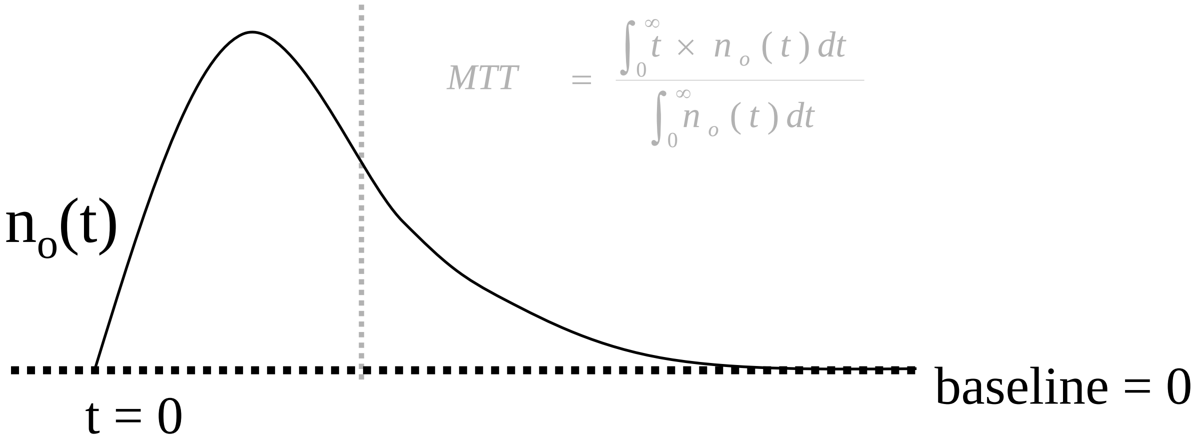




Chih-Kuang Yeh and P. C. Li, “Contrast specific ultrasonic flow measurements based on both input and output time intensities,” *Ultrasound in Medical & Biology*, 2002

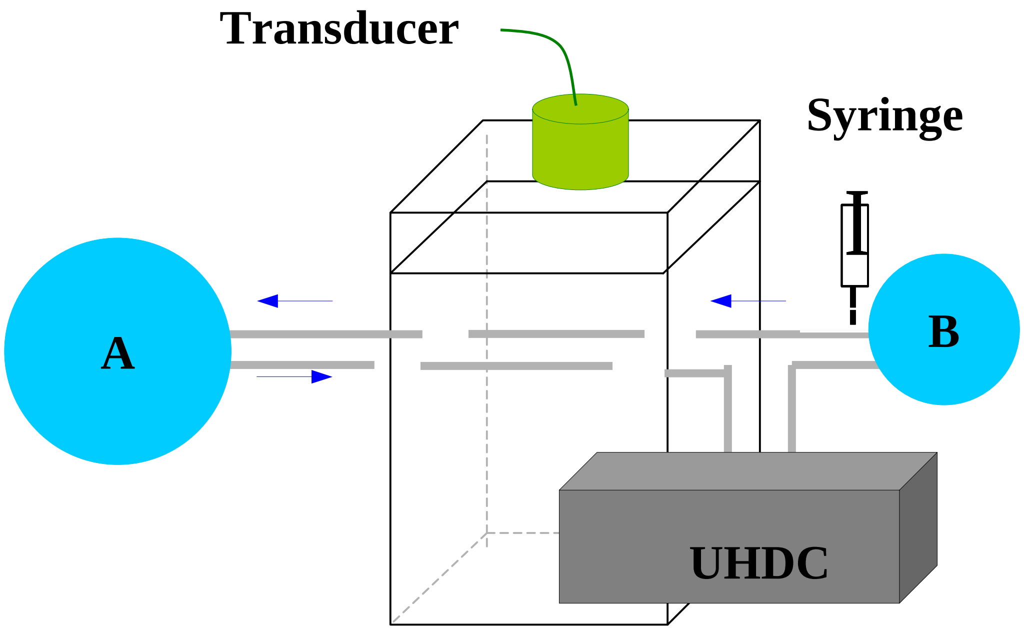
Assessment of Parameters in Pulsatile Flow using Ultrasound Contrast Agent

- Provide a model for the assessment of perfusion characteristics
- Dilution theory
  - MTT : mean transient time
  - theory : V/Q (ideal)
- LTI system ➡ TV (time-varying)



$$MTT = \frac{\int_0^\infty t \times n_o(t)\,dt}{\int_0^\infty n_o(t)\,dt}$$

Experimental setup

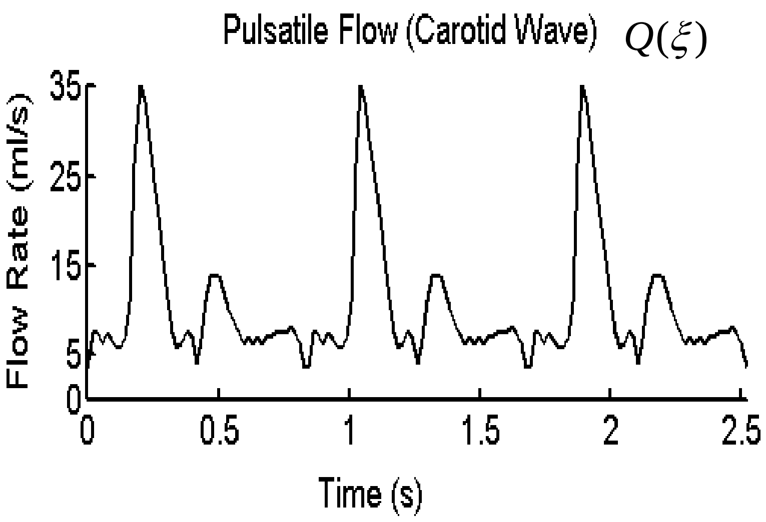


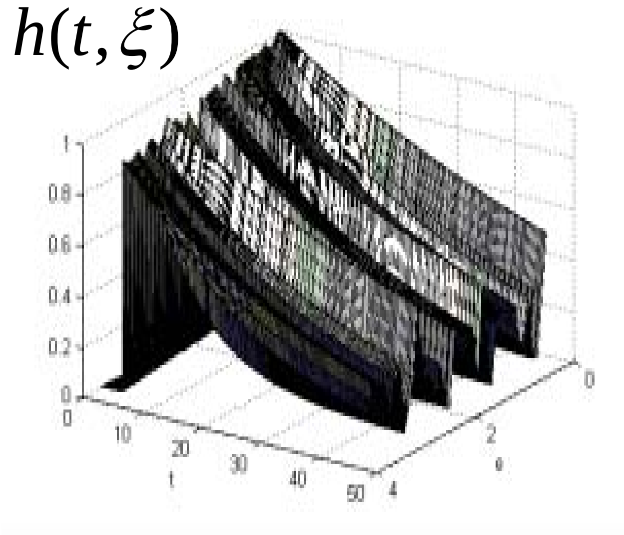
Simulation methods

Superposition theory :

$$I_o(t) = \int_{-\infty}^{\infty} h(t,\xi) I_I(\xi) d\xi$$

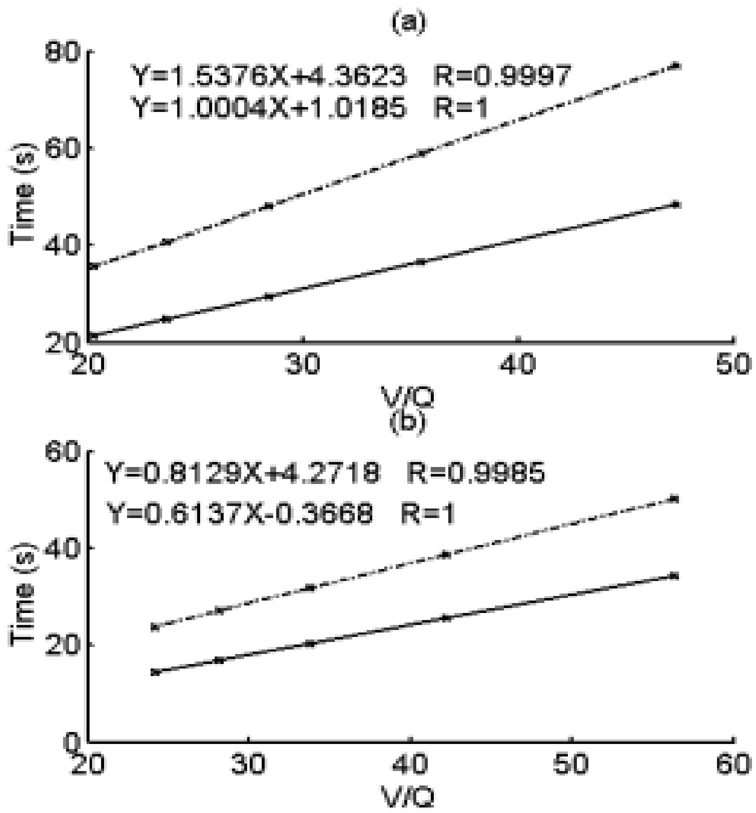
$$h(t,\xi) = \begin{cases} 0 & t < \xi \\ e^{-Q(\xi)t/V} & t \geq \xi \end{cases}$$



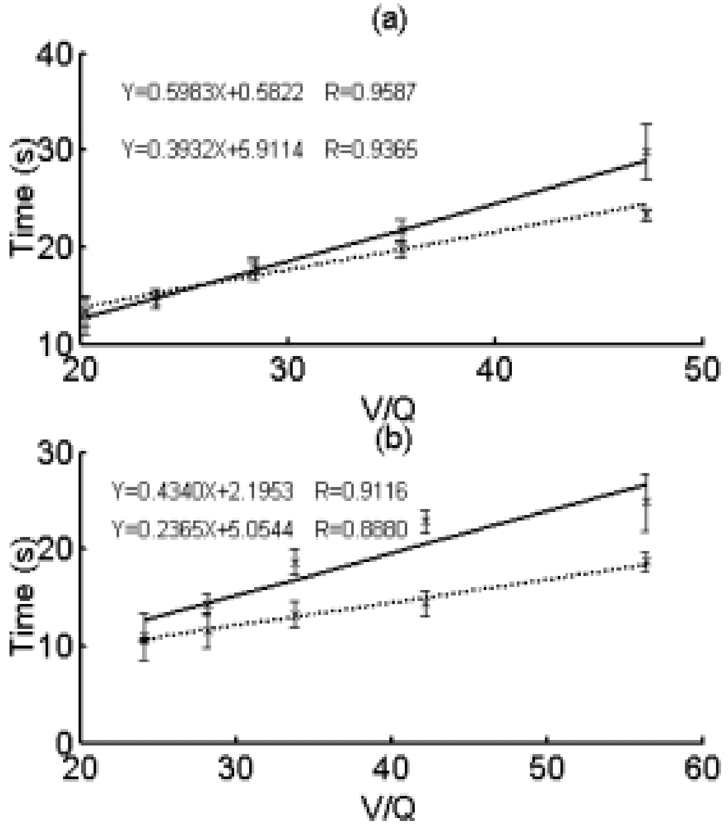


Results: the theoretical values & MTT

simulation result



experimental result



EQUIPMENT (1)

Panametrics Model 5900 PR (pulser / receiver)	1	200 MHz digital
GW Model GFG-813 (function generator)	1	
GW Model GFG-8016D (function generator)	1	
HP Model 54603B (oscilloscope)	1	60 MHz 附 probe X 4 power line X 1
Tektronix Model TDS 380 (oscilloscope)	1	Two channel / digital real-time / 400MHz / 2GS/s
TAIK Model TK-12001D (DC power supply)	1	
EPE Model EP-3000 (DC power supply)	1	
Cimarec Model SP46925 (stirrer/heater)	1	S/N:1069980822742
OHAUS Model IP12KS	1	
Microtime Model 51/52-E (WINICE)	1	S/N:A00I955172
Cole Parmer Model 07596-20 (damper)	1	
Cole Parmer Model 77021-60 (pumper)	1	
GaGe Model CompuGen 1100 (AFG)	1	ISA interface 1M RAM on board S/N: G00086 80 MHz
GaGe Model CompuScpoe 12100 (A/D)	1	PCI interface 1M RAM on board S/N: P10243 100 MHz
Amplifier Research LN1000A(LNA)	1	DC transformer
Amplifier Research P25A250A(PA)	1	
Signatec DAC200	1	User manual , CDX1,BNC to SMD cableX3
Signatec PDA500	1	User manual , CDX1
Signatec PMP8-A	1	User manual , CDX1
Panametrics HF cable	3	1 ft., 3 ft., 6ft.

EQUIPMENT (2): Transducer

EQUIPMENT (3): Commercial Ultrasound Machine

(SonoSite)
(hand-carried ultrasound system)

(GE LOGIQ500)

EQUIPMENT (4): Phantom

Breast (I)

Breast (II)

Baby

EQUIPMENT (5)

(Digital Sonifier, BRANSON)
Making Microbubbles

(UHDC Flow System)
Simulation Physical Pulsatile Flow

EQUIPMENT (6)

Degas Equipment