超音波影像實驗室
Ultrasonic Imaging Laboratory

Primary Investigator
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Ultrasonic Imaging Laboratory Members

Advisor: Dr. Pai-Chi Li
Post-doc: 1
PhD Student: 8
Master Student: 5
Research Assistant: 2
Administration Assistant: 1
Pai-Chi Li (Professor, Senior Member of IEEE Society)
Electrical Engineering
National Taiwan University
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PERSONAL EDUCATION

<table>
<thead>
<tr>
<th>Institution</th>
<th>Years</th>
<th>Degree</th>
<th>Field</th>
</tr>
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<tbody>
<tr>
<td>National Taiwan University</td>
<td>1983-1987</td>
<td>B.S.</td>
<td>Electrical Engineering</td>
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<tr>
<td>University of Michigan</td>
<td>1989-1990</td>
<td>M.S.</td>
<td>EE: Systems</td>
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<td>University of Michigan</td>
<td>1991-1994</td>
<td>Ph.D.</td>
<td>EE: Systems</td>
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RESEARCH INTEREST

Signal and imaging processing, Ultrasonic medical imaging

WORK EXPERIENCE

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<th>Years</th>
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<tr>
<td>Adjunct Associate Investigator</td>
<td>Health Research Institutes, Taiwan</td>
<td>2001-present.</td>
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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

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.

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=20 cm/s,
Doppler angle=45°,
SNR=10 dB.
Ultrasonic Small Animal Imaging
High Frequency Ultrasonic Imaging System

- Fully digital system architecture
- 50 MHz center frequency
- 60% fractional bandwidth
High Frequency Ultrasonic System: Rat

Rat B-mode kidney image

The B-mode images of embryo’s head.

High Frequency Ultrasonic System: Zebra-Fish

Color flow image of Zebra-Fish in 25 MHz

Zebra-Fish (3-5 cm)

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 (micron)

Liposomes (nano)

Not a drug vehicle

Mainly hydrophilic drug vehicle
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.

Current work:
1. Liposome (with dye Direct-81 red) as O.A. contrast gent
2. Gold nanoparticles as P.A. contrast agent
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.

Coherent: \( CF \approx 1 \)

Incoherent: \( CF < 1 \)

\[
CF = \frac{\left| \sum \frac{Rf (t - \Delta t)}{N} \right|^2}{N \sum \left| \frac{Rf (t - \Delta t)}{N} \right|^2}
\]

\[0 \leq CF \leq 1\]
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

\[ \frac{\partial \rho'}{\partial t} + \rho_0 \nabla \cdot \vec{u} = -\rho' \nabla \cdot \vec{u} - \vec{u} \cdot \nabla (\rho' + \rho_0) \]

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

\[ \rho_0 T_0 \frac{\partial s}{\partial t} = \nabla \cdot (\kappa \nabla T') + W \]

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

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

- \( \rho' \): density deviation
- \( \vec{u} \): particle velocity
- \( T' \): temperature deviation
- \( s \): entropy deviation
- \( p \): acoustic pressure
- \( W \): heat generation function
- \( \kappa \): thermal conductivity
- \( \mu \): viscousity
- \( C_p, C_v \): specific heats
- \( \rho_0, T_0 \): 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

Laser irradiation

Water Transparent

Glycerin
Absorption Coef. = 20 cm$^{-1}$

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 than 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
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 ($f_x$)

No aberration

Aberrated

10 dB/div

More high frequency components
Generalized Coherence Factor

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

- 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.

No aberration

Aberrated
Results

Simulation

Aberrated

GCF Corrected

Experiment

Anechoic cyst
2. Ultrasonic Nonlinear Imaging
Ultrasonic Tissue Harmonic Imaging

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

  ![Diagram of Pressure before and after distortion](image)

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

  ![Diagram of Positive and Negative pulse](image)
Ultrasonic Tissue Harmonic Imaging

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

*Multi-focus technique*

*Secondary focus*

*Primary focus*

*harmonic beam pattern*

![Graph showing velocity vs depth and harmonic beam pattern](image-url)
Ultrasonic Tissue Harmonic Imaging

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

Diagram showing:
- Image Plane
- Single Crystal Transmitter
- Array Receiver
- Speckle Target

Graphs showing correlation coefficient versus normalized distance for different SNR values:
- 2MHz Fundamental
- 4MHz Second Harmonic
- 3.5MHz Fundamental
- 7MHz Harmonic
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**
  \[
  X^{m-1}(i, j) = X^m(2i-1, 2j-1) + X^m(2i-1, 2j) + X^m(2i, 2j-1) + X^m(2i, 2j)
  \]
  \[
  SAD^m(X, Y) = \sum_{i=1}^{2^m} \sum_{j=1}^{2^m} |X^m(i, j) - Y^m(i, j)|
  \]

- **Multilevel Block matching**
  - Reduce *numbers* of points to be searched
  - We use 2 levels
  - 1st level: window size= \(w \times w\), 9 points
  - 2nd level: window size= \(\frac{1}{2} \times w \times \frac{1}{2} \times w\), all points
Results

- Algorithm performance For 121 pixels

<table>
<thead>
<tr>
<th>Language</th>
<th>FSA (s)</th>
<th>BSPA (s)</th>
<th>BSPA &amp; Multilevel (s)</th>
<th>Ratio</th>
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</thead>
<tbody>
<tr>
<td>Matlab</td>
<td>27.14</td>
<td>14.17</td>
<td>7.64</td>
<td>3.6 : 2 : 1</td>
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<tr>
<td>C</td>
<td>12.3</td>
<td>1.53</td>
<td>0.998</td>
<td>12 : 1.5 : 1</td>
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</table>

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

- Compounding image

Liver
SNR=0.1164
Computing speed ↑

L: original image
R: compounding image with BSP & Multilevel
Cont’d

Thyroid
SNR=0.9625
SNR ↑
Computing speed ↑

Breast
SNR=0.1692
SNR ↑
Computing speed ↑

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

\[ \text{Slope (Young's modulus)} = 1701.3 \text{ Pascals} \]
The Young’s modulus of normal liver, cirrhotic liver and hepatic tumors

<table>
<thead>
<tr>
<th>Preload strain</th>
<th>Young’s Modulus (Pascals)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Liver parenchyma (mean value)</td>
</tr>
<tr>
<td></td>
<td>Normal (Fibrosis score: 4)</td>
</tr>
<tr>
<td>5%</td>
<td>642.6</td>
</tr>
<tr>
<td>10%</td>
<td>1083.6</td>
</tr>
</tbody>
</table>
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

1. Acquire 2 images
2. Speckle tracking
3. Find $x'$, $z'$, and $\theta'$
4. Correct the $x'$ and $z'$ motions
5. Images are spatial matched
6. Calculate 2D C.C.
7. Go through all $\beta$, $\gamma$, $Y$ combinations
8. Find the best matched combination

$x', y', z', \theta', \phi', \psi'$
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

Hardware Design

Dynamic focus

Tissue attenuation

Lateral filter

Demod.

Axial filter

ADC

Coded excitation

DAC

Tissue attenuation

PR5900

0.4xGCPuls

0.6xGCPuls

Intensity (dB)

Frequency (MHz)
Wire phantom

- 52µm nylon wire phantom
- Gaussian Pulse
- Gaussian Chirp Pulse
Resolution test

- Lateral projection:
- Spatial resolution is about 60 µm

![Graph showing intensity vs position and depth for GPS, GCPS, and AFT PC]
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

\[ \hat{f} = \frac{\theta(T)}{2\pi T}, \quad T : PRI \]
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)

Experimental Flow Result

Traditional Butterfly Search Line

Multiple Butterfly Search Lines

Traditional RF Butterfly

Multi-line Butterfly
High Frequency Flow Estimation

Color Flow Image
High Frequency Harmonic Image

25MHz 300 µm Cyst image

Fundamental Image

Harmonic Image

Fundamental Image

Harmonic Image

Frequency (MHz)

dB

0  10  20  30  40  50  60  70  80  90  100
Doppler Blood Flow Estimation (I)

- Doppler Angle Estimation
- Blood Flow Estimation


Blood Flow Estimation Using Ultrasonic Contrast Agent

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

Doppler Blood Flow Estimation in Pulsatile Flow (II)

- Doppler Angle Estimation
- Blood Flow Estimation

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 $\rightarrow$ TV (time-varying)

\[
\text{MTT} = \frac{\int_0^\infty t \times n_o(t) \, dt}{\int_0^\infty n_o(t) \, dt}
\]
Experimental setup
Simulation methods

Superposition theory:

\[ I_o(t) = \int_{-\infty}^{\infty} h(t, \xi) I_1(\xi) d\xi \]

\[ h(t, \xi) = \begin{cases} 
0 & t < \xi \\
 e^{-Q(\xi) t/V} & t \geq \xi 
\end{cases} \]
Results: the theoretical values & MTT simulation result

Experimental result
# EQUIPMENT (1)

<table>
<thead>
<tr>
<th>Equipment Name</th>
<th>Quantity</th>
<th>Specifications</th>
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</thead>
<tbody>
<tr>
<td>Panametrics Model 5900 PR (pulser / receiver)</td>
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<td>200 MHz digital</td>
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<tr>
<td>GW Model GFG-813 (function generator)</td>
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<tr>
<td>GW Model GFG-8016D (function generator)</td>
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<td>HP Model 54603B (oscilloscope)</td>
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<td>60 MHz 配 序 X 4  power line X 1</td>
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<td>Tektronix Model TDS 380 (oscilloscope)</td>
<td>1</td>
<td>Two channel / digital real-time / 400MHz / 2GS/s</td>
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<td>TAIK Model TK-12001D (DC power supply)</td>
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<td>EPE Model EP-3000 (DC power supply)</td>
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<td>Cimarec Model SP46925 (stirrer/heater)</td>
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<td>OHAUS Model IP12KS</td>
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<td>Panametrics HF cable</td>
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<td>1 ft., 3 ft., 6ft.</td>
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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