Basic Principles - 2D Echo and Doppler

Joe M. Moody, Jr, MD
UTHSCSA and STVHCS
Formatted 2008

Acknowledging many illustrations from Weyman’s text and others.
Echo-Doppler Basic Principles

• Ultrasound physics
  – resolution - axial and lateral
  – attenuation

• Doppler ultrasound
  – Aliasing
  – Bernoulli principle
  – laminar and turbulent flow
  – continuity equation
Ultrasound physics

- Sound - waves of compression and rarefaction propagated through a medium
- Ultrasound - sound frequency above 20kHz
- Cardiac ultrasound - 1 - 20 mHz
- Intensity - watts/cm² (joule/sec/cm²)
Wavelength times frequency equals propagation velocity:
\[ c = \lambda \times f \]
and
\[ c = 1540 \text{ m/s}, \] so
\[ \lambda \text{ (mm)} = \frac{1.54}{f} \text{ (MHz)} \]

- \( \lambda \) for 2 MHz is 0.75 mm
- \( \lambda \) for 4 MHz is 0.38 mm
Piezoelectricity

- Mechanical stress applied to a crystal causes electrical charges bound in the crystal to shift to the surface where they can be measured as a voltage.
- Electric current applied to a crystal changes the crystal shape, alternating current can cause vibration of crystal, producing sound wave.
Transducer Structure

Wavelength versus Penetration

Echo Transducer

- Definitions of fields of transducer performance
Echo Transducer

- Single crystal
- Effect of crystal diameter and frequency on near field and far field
- Larger diameter and higher frequency give longer near field and less divergent far field
Transducer Beam Zones

Echo Transducer

- Effect of focal length and focusing on near field and far field
Graph: Transducer Beam Zones

Solid line: length of near zone
Dashed line: divergence angle

Side Lobe Artifacts
Side Lobe Artifacts
Echo Transducer Lateral Variation
Side Lobe Artifact in Single Crystal Transducer

Position of side lobes at locations where the distances from each edge of the crystal face differ by one wavelength

Position of grating lobes is determined by spacing between centers of independent crystal elements in the transducer.

\[ S = \text{spacing between elements} \]

\[ F = \text{focal length} \]

\[ \lambda = \text{wavelength} \]

Echo Transducer Axial Variation
Phased Array Echo Transducer
Phased Array Echo Transducer

- Effect of electronic focus
Destiny of Sound Wave

### Velocity of Sound in Air and Various Tissues

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Velocity (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>330</td>
</tr>
<tr>
<td>Fat</td>
<td>1450</td>
</tr>
<tr>
<td>Water</td>
<td>1480</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>1540</td>
</tr>
<tr>
<td>Kidney</td>
<td>1560</td>
</tr>
<tr>
<td>Blood</td>
<td>1570</td>
</tr>
<tr>
<td>Muscle</td>
<td>1580</td>
</tr>
<tr>
<td>Bone</td>
<td>4080</td>
</tr>
</tbody>
</table>

Clinically, use 1500 m/s

Velocity and Time Relation in Echocardiography

- **Distance** = rate * time
- **Time** = distance/rate
- **Distance/rate** (for 15 cm) =
  - $0.15\text{m}/1500\text{m/sec} = 0.0001\text{ sec, or } 1/10,000\text{ sec for one way trip}$
  - $0.0002\text{ sec or } 2/10,000\text{ sec for 2 way trip}$
  - $0.0003\text{ or } 3/10,000\text{ sec for 2 way trip of 20 cm}$
Display of Echo Signal

A = amplitude  B = brightness  M = motion
Echo Interfaces

- Velocity of sound in a medium depends on density (denser is faster) and elasticity of the medium
- Human tissue - 1540 m/sec, faster in bone
- Acoustic mismatch or change in acoustic impedance of an interface causes a reflection
- Interface perpendicular to beam is strongest
Echo Reflection

• **Specular reflection** - reflector is large and smooth relative to ultrasound wavelength - responsible for the echo images, angle of incidence is important

• **Scattered reflection** - reflector is small and rough relative to ultrasound wavelength - responsible for some images and critical for Doppler
Echo Resolution and Attenuation

- **Axial resolution** - better with higher frequency and fewer cycles/pulse (packet size) 3.5MHz=0.43mm wavelength
- **Lateral resolution** - varies with transducer size, shape, frequency, and focusing
- **Attenuation** - worse with high frequency
- **Attenuation** - half-value layer (35cm in blood, 3.6 cm in muscle)
## Echo Attenuation – Half-power Distance

<table>
<thead>
<tr>
<th>Tissue Type</th>
<th>Distance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>380 cm</td>
</tr>
<tr>
<td>Blood</td>
<td>15 cm</td>
</tr>
<tr>
<td>Soft tissue (except muscle)</td>
<td>1-5 cm</td>
</tr>
<tr>
<td>Muscle</td>
<td>0.6-1 cm</td>
</tr>
<tr>
<td>Bone</td>
<td>0.2-0.7 cm</td>
</tr>
<tr>
<td>Air</td>
<td>0.08 cm</td>
</tr>
<tr>
<td>Lung</td>
<td>0.05 cm</td>
</tr>
</tbody>
</table>

Axial Resolution

- Better with:
  - high frequency
  - short packet length
Time Constraint in Echo-Doppler

- 1540 m/sec in tissue
- 20 cm depth is 40 cm round trip
- 3850 round trips/sec (M-mode)
- 150 round trips for one image
- 25 images/sec
- Trade-off: temporal resolution, spatial resolution (line density) and depth

\[ P.D. = \frac{D}{C} = \frac{15\text{cm}}{1.5\text{mm/\mu sec}} = 100\mu\text{sec (200)} \]

\[ PRF = \frac{1}{0.002 \text{ sec/pulse or 5000 pulses/sec}} \]
Time Constraint in Echo-Doppler

- 1540 m/sec in tissue, 20 cm depth is 40 cm round trip, 3850 round trips/sec (M-mode), 150 round trips for one image, 25 images/sec
- Trade-off: temporal resolution, spatial resolution (line density) and depth
- In the “Res” mode, there is improved temporal resolution
Time Constraint in Echo-Doppler

1. 

\[
\begin{align*}
4500 \\
\text{d} = 30^\circ \times 30
\end{align*}
\]

\[
\begin{align*}
4500 \\
\text{d} = 5 \text{ lines/degree}
\end{align*}
\]

2. 

\[
\begin{align*}
4500 \\
\text{d} = 150 \text{ lines/field}
\end{align*}
\]

\[
\begin{align*}
4500 \\
\text{d} = 1.7 \text{ lines/degree}
\end{align*}
\]

3. 

\[
\begin{align*}
4500 \\
\text{d} = 90^\circ \times 30
\end{align*}
\]
The larger packet size, the better Doppler discrimination, the worse the echo discrimination.
Time Constraint in Echo-Doppler

Table 11–2. Typical Combinations of Depth of Field, Packet Size, Lines per Frame, and Frame Rate for Color Flow Images

<table>
<thead>
<tr>
<th>Depth of Field (cm)</th>
<th>Packet Size</th>
<th>Lines per Frame</th>
<th>Frames per Second</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>4</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>8</td>
<td>8</td>
<td>45</td>
<td>20</td>
</tr>
<tr>
<td>10</td>
<td>8</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>12</td>
<td>4</td>
<td>45</td>
<td>30</td>
</tr>
<tr>
<td>14</td>
<td>8</td>
<td>30</td>
<td>20</td>
</tr>
<tr>
<td>16</td>
<td>8</td>
<td>45</td>
<td>15</td>
</tr>
<tr>
<td>18</td>
<td>4</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>18</td>
<td>4</td>
<td>45</td>
<td>20</td>
</tr>
<tr>
<td>18</td>
<td>8</td>
<td>45</td>
<td>12</td>
</tr>
</tbody>
</table>

\[ F_0 = 2.5 \text{ MHz} \]
\[ \text{PRF} = 5 \text{ KHz} \]
Artifact From Scatter

Artifact From Scatter

Refraction of ultrasound beam by intervening tissue

double-image of Ao

transducer

specular reflector

LV

Breakthrough: Harmonic Imaging

• Since about 1997 with the introduction of harmonic imaging, there has been a dramatic improvement in image quality.

• Today, essentially all images are obtained with harmonic imaging.

• Sound travels a little faster at the peak of the sound wave (more compressed) than the trough, so with each subsequent waveform a small amount of harmonic is generated (1962), similar to the breaking of the crest of a wave at the beach.

2 Types of Harmonic Imaging

- Harmonic energy in reflection can occur with echo-contrast agents which resonate with ultrasound stimulation and produce harmonic emission.
- Harmonic energy in transmission occurs due to the compressibility of the tissue.

Harmonic Imaging

Classical ultrasound theory:
• Energy propagation is linear
• Different frequencies travel at the same speed in the same medium
• New frequencies should not appear
• Attenuation only reduces amplitude

Harmonics:
• Some objects in the path of the beam may resonate and emit higher frequencies
• Transmission of ultrasound through a compressible medium yields harmonics
• Signal grows with distance
• Signal generation is nonlinear
• Reduced backscatter

Harmonic Imaging

- The reflected pulse has harmonic information

Harmonic Imaging

- The propagated pulse is distorted by the addition of harmonic energy

Harmonic Imaging

- The harmonic energy increases with depth

• Strength of harmonics increases as square of source energy, so weak reflections produce poor harmonics

Harmonic Imaging

- Boosting of harmonic information is required

Harmonic Imaging

- Narrow band pulse is smoother in initiation and termination

Harmonic Imaging

- The square pulse (wide band) has significant energy at many frequencies, including at the harmonic frequency.
- In contrast, the smooth pulse (narrow band) has almost no energy at the harmonic frequency.

Harmonic Imaging

- Narrow band pulse allows filtered signal to be essentially free of fundamental frequency reflection

New Instrumentation: High PRF Equipment

• The classical time constraints explained earlier are rendered nonconstraining by the process of multiple simultaneous scan line analysis

• I don’t understand it
Doppler Cardiography

- Primary source of ultrasonic reflection - RBC
- Scattered reflector
- Motion with respect to transducer causes shift in frequency of sound wave, the measurement of which is fundamental to the Doppler signal
Doppler Shift

Doppler Shift

Doppler shift \( = (F_s - F_T) \)

\( F_T \) is frequency transmitted, and \( F_S \) is scattered frequency received

\[ v = \frac{c \ (F_s - F_T)}{2 \ F_T (\cos \Theta)} \]

\( V \) is velocity of blood
\( C \) is speed of sound in blood (1540 m/s)
\( \Theta \) is intercept angle between beam and blood flow
\( 2 \) is factor to correct for 2 trips

Doppler Equation

\[ v = \frac{c (F_S - F_T)}{2 F_T \cos \theta} \]

- \( v \) is velocity of blood
- \( C \) is speed of sound in blood (1540 m/s)
- \( F_S - F_T \) is Doppler shift
- \( \Theta \) is intercept angle between beam and blood flow
- 2 is factor to correct for 2 trips

\[
V = \frac{1540 \text{ m/s} \times (2.009 - 2.000) \text{ MHz}}{2 \times 2.000 \text{ MHz} (1.0)} = 1540 \times 0.009 / 4 = 3.465 \text{ m/sec}
\]

Doppler Equation

\[ v = \frac{c (F_s - F_T)}{2 F_T (\cos \theta)} \]

V is velocity of blood
C is speed of sound in blood (1540 m/s)
\( F_s - F_T \) is Doppler shift
\( \Theta \) is intercept angle between beam and blood flow
2 is factor to correct for 2 trips

Ignoring the speed of sound and cosine components, a direct relationship exists between Doppler shift & transmit frequency & blood velocity:
1.3 KHz shift for 1.0 MHz per 1.0 m/sec velocity
2.6 KHz shift for 1.0 MHz per 2.0 m/sec velocity
2.6 KHz shift for 2.0 MHz per 1.0 m/sec velocity
5.2 KHz shift for 2.0 MHz per 2.0 m/sec velocity

Timing in Pulsed Doppler: The PRF

PRF is mainly limited by depth in Pulsed Doppler Ultrasound.

**Pulse Cycle Consists of Three periods:**
- Transmission (duration affects velocity resolution)
- Travel time (duration determines depth)
- Reception (duration determines sample volume)

A waveform must be sampled at least twice in each cycle for accurate determination of wavelength. Therefore, the maximum detectable frequency shift (the Nyquist Limit) is one-half the PRF. But the maximal detectable velocity depends on the equation.

Nyquist Limit

• The maximum detectable frequency shift (the Nyquist Limit) is one-half the PRF. But the maximal detectable velocity depends on the equation.

\[ v = \frac{c \left( F_s - F_T \right)}{2 F_T \cos \theta} \]

Let’s say the PRF is 5000.
Nyquist limit = (2500Hz or 2.5KHz = .0025MHz) = 
\[ \frac{1540 \text{m/s} \times (\text{PRF}/2) \text{MHz}}{2 \times 2.000 \text{MHz}(1.0)} = \]
\[ \frac{1540 \text{m/s} \times (0.0025) \text{MHz}}{2 \times 2.000 \text{MHz}(1.0)} = \]
1540 \* 0.00025/4 = 0.9625 m/sec = 96cm/sec
High PRF Doppler

Color Doppler Diagram

Color Doppler Diagram

Type of Doppler Signal

- Pulsed wave Doppler
- Continuous wave Doppler
- Color flow Doppler (a form of pulsed wave Doppler)
- (High pulse repetition frequency pulsed wave Doppler is intermediate between pulsed and continuous wave)
### Types of Doppler Signal

<table>
<thead>
<tr>
<th></th>
<th>Pulsed wave</th>
<th>Continuous wave</th>
<th>High PRF</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Aliasing</strong></td>
<td>Yes</td>
<td>No</td>
<td>Yes, better</td>
</tr>
<tr>
<td><strong>High velocity</strong></td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Range specific</strong></td>
<td>Yes</td>
<td>No</td>
<td>Some ambiguity</td>
</tr>
<tr>
<td><strong>Laminar Resolution</strong></td>
<td>Yes</td>
<td>No</td>
<td>Yes, somewhat</td>
</tr>
</tbody>
</table>
Principles of Imaging

• Reducing scan angle can increase line density
• Side lobe artifacts can be generated by the transducer
• Signal processing can alter relationship between strength of received signal and display strength
Principles of Imaging - 2

• Persistence of image on screen smoothes discontinuities (temporal image processing)
• Signal processing is complex
• Connection with video monitor is not trivial, and can result in signal loss
Principles of Imaging

- Persistence of image on screen smoothes discontinuities (temporal image processing)
- Signal processing is complex
- Connection with video monitor is not trivial, and can result in signal loss

Big Otto, 1997
Principles of Imaging - Contrast

• Contrast enhancement with specular reflectors
  – air microbubbles
  – protein microparticles

• Contrast location
  – bloodstream
  – myocardial (experimental)
Imaging Advances

• Imaging traditionally looks for the frequency transmitted
  – Human tissue naturally causes increase in frequency of returned signal, and can be imaged at twice the transmitted frequency
  – Native tissue harmonic imaging

• Tissue characterization
Echo Artifacts

- Incorrect persistence on screen - bright object may last into subsequent frames
- Point spread function in far field
- Internal reverberations, projected at twice the real distance
- Reverberations from highly reflective interface may be a series of echoes
- Shadowing behind a strong reflector
Uses of Doppler Information

• Analysis of velocity
• Analysis of turbulence
• Analysis of valve area
  – MV - pressure half time
  – Continuity Equation
• Analysis of pressure difference, instantaneous or mean
Blood Flow

- The original and still most commonly used source of Doppler information
- Flow is generally either laminar or turbulent depending on whether Reynold’s number is exceeded (about 5,000-10,000)
- Reynolds: \(2RVp/n\) where \(R=\)radius, \(V=\)velocity, \(p=\)density, \(n=\)viscosity
- Flow is usually at least somewhat pulsatile

Doppler Pressure Gradient

\[ \Delta P = \frac{1}{2} \rho (v_2^2 - v_1^2) + \rho \int_1^2 \frac{d\vec{v}}{dt} + R(\vec{v}) \]

Convective acceleration plus flow acceleration plus viscous forces

Bernoulli Equation

\[ \Delta P = 1/2 \rho (v_2^2 - v_1^2) + \rho \int_1^2 \frac{\vec{d} \vec{v}}{dtds} + R(\vec{v}) \]

1. **Convective acceleration** – Velocity squared
   
   Pressure energy $\rightarrow$ kinetic energy

2. **Flow acceleration** – Derivative of velocity
   
   Energy to impart momentum

3. **Viscous forces** – Velocity
   
   Energy losses from friction between neighboring fluid elements, more with turbulence

3.972 $\approx$ 4
Bernoulli Equation

1. **Convective acceleration**
2. **Flow acceleration**
3. **Viscous forces**

\[ \Delta P = \frac{1}{2} \rho (v_2^2 - v_1^2) + \rho \int_{1}^{2} \frac{d\vec{v}}{dt} \, ds + R(\vec{v}) \]
Doppler Pressure Gradient

- At peak velocity, flow acceleration is zero.
- Viscous forces are negligible when flow is high and orifice is small.
- So, the pressure gradient is by convective acceleration alone, and by substituting appropriate constants and neglecting proximal velocity, $\Delta P = 4v^2$. 
Pulsed Doppler Limit

• Nyquist limit: Aliasing occurs when the frequency of the Doppler shift exceeds 1/2 the PRF
  – Doppler shift is proportional to transducer frequency
  – More with greater depth of sample volume
  – Decrease effect by shift of baseline
  – Decrease effect by increasing angle of signal
Doppler Artifacts

- Aliasing (also range ambiguity)
- Mirroring - when Doppler shift is displayed as equal frequency and opposite direction (solve by decreasing gain)
- Display of external audible noise as Doppler
- Signal loss by data sharing
- Beam width artifact
Color Doppler Artifacts

- Color aliasing
- Reverberations
- Effects of wall shadowing
  - usual suppression
  - suppression by strong echo signal
- Effects of flow angle in the scan plane
Color Doppler M-Mode

From GE website
Doppler Advances

• Contrast Doppler for myocardial perfusion
  – Intracoronary
  – Intravenous (perfluorocarbons)

• Doppler analysis of tissue
  – Wall motion
  – Strain
Tissue Doppler Background

• Signal from tissue is different than that from blood pool
  – Blood flow is 10 times faster than wall motion
  – Blood flow signal is much weaker (40dB) than wall motion signal
Tissue Doppler Imaging Techniques

- High pass wall filter is disabled
- Gain amplification for low velocity or reduction for myocardium
- Expanded scale (peak less than 25 cm/s)
- Small (2 mm) sample volume ("gate")

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Gate: lateral LV base in A4C
- V1 – systolic
- V2 – early diastolic
- V – late diastolic

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Gate: midventricular septum in A4C
- Large gate (20mm) gives inadequate recording

Waggoner AD et al. J Am Soc Echocardiogr 2001;
Table 1 Advantages and limitations of color tissue Doppler imaging (TDI) versus pulsed wave (PW) TDI

<table>
<thead>
<tr>
<th></th>
<th>Advantages</th>
<th>Limitations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Color TDI</td>
<td>• Spatial orientation of myocardial velocities can be seen superimposed on the real-time 2-dimensional image</td>
<td>• Poor temporal resolution caused by longer processing times involved with autocorrelation analysis</td>
</tr>
<tr>
<td></td>
<td>• Myocardial velocities can be displayed in a manner similar to that of conventional color flow imaging</td>
<td>• Typically requires off-line analysis for quantification of the myocardial velocity color maps</td>
</tr>
<tr>
<td></td>
<td>• Representation of mean myocardial velocity</td>
<td></td>
</tr>
<tr>
<td>PW TDI</td>
<td>• Real-time velocity interrogation with improved temporal resolution</td>
<td>• Only regional quantification of myocardial velocities can be done at selected sites reducing spatial resolution</td>
</tr>
<tr>
<td></td>
<td>• Ability to quantitate peak rather than mean myocardial velocities</td>
<td>• Sampling cannot be localized to the endocardial or epicardial layers</td>
</tr>
<tr>
<td></td>
<td>• Does not require off-line analysis</td>
<td>• Alignment of the beam parallel to the heart muscle movement may be difficult in some patients</td>
</tr>
<tr>
<td></td>
<td>• Provides instantaneous temporal display of the Doppler spectral information</td>
<td>• No correction for normal cardiac translation and rotation during sampling</td>
</tr>
<tr>
<td></td>
<td>• Objective assessment of regional function, which is especially useful in dobutamine stress echocardiography</td>
<td></td>
</tr>
</tbody>
</table>
## Tissue Doppler Imaging

**Table 2** Normal values in cm/s ± 1 SD for the basal segments of the left ventricle using pulsed TDI\(^9,23,29,41,44\)

<table>
<thead>
<tr>
<th></th>
<th>Sm</th>
<th>Em</th>
<th>Am</th>
<th>Em/Am velocity ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lateral</td>
<td>10.6 ± 2.3</td>
<td>13.3 ± 3.3</td>
<td>11.3 ± 2.9</td>
<td>1.5 ± .6</td>
</tr>
<tr>
<td>Septal</td>
<td>9.9 ± 1.7</td>
<td>11.5 ± 2.6</td>
<td>9.5 ± 2.4</td>
<td>1.0 ± .7</td>
</tr>
<tr>
<td>Anterior</td>
<td>9.2 ± 1.8</td>
<td>11.7 ± 3.4</td>
<td>10.3 ± 2.9</td>
<td>1.2 ± .7</td>
</tr>
<tr>
<td>Posterior</td>
<td>10.4 ± 2.5</td>
<td>14.3 ± 3.6</td>
<td>11.6 ± 2.6</td>
<td>1.3 ± .7</td>
</tr>
</tbody>
</table>

*Am*, Late diastolic myocardial velocity; *Em*, early diastolic myocardial velocity; *Sm*, systolic myocardial velocity.
Normal Tissue Doppler Imaging

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Normal tissue Doppler signal from basal lateral wall in A4C

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Patient with prior anteroseptal MI
  - Top: septal base
  - Bottom: lateral base
- V1 systolic velocity
- V2 diastolic velocity

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Top: rest
- Bottom: exercise
- Increases in systolic and diastolic velocities

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143
Tissue Doppler Imaging

- Htn and LVH
- Pseudonormalized LVIT pattern
- Abnormal TDI of lateral base with decreased early diastolic velocity and normal late diastolic velocity

Waggoner AD et al. *J Am Soc Echocardiogr* 2001; 14:1143
Color Tissue Doppler Imaging

b = regional isovolumic contraction time

c = regional isovolumic relaxation time

s = systole

E = rapid filling

A = atrial contraction

Trambaiolo P et al. JASE 2001; 14:85
Strain Rate Imaging

Septal wall
More yellow in the mid-anterior segment indicates increase in contractility, indicating viability

Hoffmann R et al.  JACC 2002;39:443
Tissue Doppler Applications

- Not widely used
- Diastolic function
- Regional systolic function
- Strain measurement using tissue Doppler displayed as tissue velocity imaging in color

For images to view and clips to see, try GE Medical Systems website http://www.gemedicalsystems.com/rad/us/products/vivid_7/msuvivid7img.html