Color-Doppler imaging

MEDT8002 lecture
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Background / A brief history lesson
Ultrasound imaging of blood and tissue movement

One of the main advantages of ultrasound imaging is the ability to measure and visualize the movement of tissue and blood.

Velocity mapping of carotid bifurcation flow  Velocity mapping of cardiac contractions
**CW- / PW-Doppler**

- Measures the **full spectrum of velocities** within a region of interest
- Allows for in-depth analysis of blood flow characteristics
Color-Doppler imaging

- Measures the **mean velocity** and **direction along the beam** in a 2 or 3-dimensional region of interest.
- Allows for easier detection of abnormalities manifested in blood flow patterns.

2-D CDI of a carotid bifurcation

3-D CDI of a mitral regurgitation jet
Clinical examples

Mitral valve regurgitation jet

Atrial septum defect shunt flow
More clinical examples

Carotid artery stenosis

Thyroid nodule vascularization
Doppler imaging history at NTNU

- Successful Doppler imaging instruments and its clinical use was developed in Trondheim in the late-seventies

1976: PEDOF Doppler instrument

Liv Hatle pioneered the use of Doppler ultrasound in the clinic

Bjørn Angelsen and Kjell Kristoffersen
A brief history lesson

- Technology and research progressed from single-range gate to multi-range gated Doppler, and further to 2-D Doppler imaging

Doppler imaging history at NTNU

- The company formed based on this research, now called **GE Vingmed Ultrasound**, is currently one of the world’s leading manufacturers of cardiovascular ultrasound imaging systems.

1986: Vingmed CFM 700

2009: GE Vingmed Vivid E9
A brief history lesson

- Real-time CFI was first commercially available in the mid-eighties

A brief history lesson

- However, real-time processing and display of color-Doppler images in weather RADAR was available ten years earlier (!)

*Source: G. R. Gray et al, Real-time color-Doppler RADAR display, *Bulletin of the American Meteorological Society*, vol. 56(6), 1975*
Color-Doppler processing
CDI processing blocks

- CDI data acquisition
  - Scanning operation and pulse sequence
- Clutter filtering (wall filtering)
  - Attenuating interfering signal from (near) stationary tissue
- Doppler parameter estimation
  - Estimation of Doppler power, mean-velocity, and velocity spread
- Display
  - Color encoding of Doppler parameters
What do we need to acquire in order to measure velocities in CFI?

- **Color-Doppler** is acquired using a multi-pulse scheme
  - Signal changes between pulses allows us to compute the scatterer movement

- **The number of samples available for processing (packet size)**
  - Cardiac imaging: 8-10
  - Vascular imaging: 10-16
  - Abdominal imaging: 10-12

Scan sequence in CDI acquisition
(PRF = Pulse repetition frequency)
Data acquisition in CDI

Mechanical scanning
+ No settling time for clutter filter
- Low frame rate

Electronic packet scanning
- Settling time for clutter filter
+ Flexible PRF without loss in frame rate

Electronic continuous scanning
+ No settling time for clutter filter
- High frame rate, but low PRF
Extracting the blood signal

Clutter filtering

- The Doppler spectrum may consist of three components, clutter $c$, blood $b$, and random noise $n$
- The clutter is much stronger than blood and must be attenuated
- Blood typically has a higher velocity than tissue, i.e. higher Doppler shifts → Attenuate lower velocities
General clutter filter design

- Clutter filters should have a high stop-band attenuation (60-80 dB) to sufficiently attenuate clutter
- Clutter filters should have a short transition region to avoid removing signal from blood
Clutter filtering challenge

- If we had an infinite amount of samples available and a stationary process, close to ideal filter could be designed

- Due to 1) flow dynamics, and 2) the limited acquisition rate in ultrasound, the number of samples available for filtering is very limited

- Only 8-16 samples => nonideal filters!
  - False coloring of tissue regions
  - Flow signal is also attenuated
Estimating blood velocity

- The parameters mean power $P$, mean velocity $v$, and velocity spread $B$ are estimated.
- Both temporal and spatial smoothing is used to achieve robust measurements.

![Doppler velocity spectrum diagram](image)
Pulsed Doppler signal model
Doppler parameter estimation

Doppler parameter estimation in CFI has focused on the first three moments of the Doppler spectrum, which equals the mean power, mean frequency, and bandwidth (rms):

\[
P = \int_{-\infty}^{\infty} G(\omega) \, d\omega \quad \bar{\omega} = \frac{\int_{-\infty}^{\infty} \omega \cdot G(\omega) \, d\omega}{\int_{-\infty}^{\infty} G(\omega) \, d\omega} \quad B_{\text{rms}}^2 = \frac{\int_{-\infty}^{\infty} (\omega - \bar{\omega})^2 \cdot G(\omega) \, d\omega}{\int_{-\infty}^{\infty} G(\omega) \, d\omega}
\]

However: estimating the Doppler power spectrum and integrating is not a practical solution.

Time (phase) domain approaches has several qualities
- Less computationally expensive
- Robust in low signal-to-noise ratios
- Velocity range covering the full Nyquist spectrum width
Time domain formulation

The *Wiener-Khinchin* formula relates the autocorrelation function and the power spectral density function:

\[
R(\tau) = \frac{1}{2\pi} \int_{-\infty}^{\infty} G(\omega) e^{j\omega \tau} d\omega
\]

Derivatives with respect to \( \tau \) gives:

\[
\dot{R}(\tau) = \frac{j}{2\pi} \int_{-\infty}^{\infty} \omega G(\omega) e^{j\omega \tau} d\omega
\]

\[
\ddot{R}(\tau) = \frac{-1}{2\pi} \int_{-\infty}^{\infty} \omega^2 G(\omega) e^{j\omega \tau} d\omega
\]

Yields time-domain expressions for power, mean frequency, and bandwidth (rms):

\[
P = R(0) \quad \bar{\omega} = -j \frac{\dot{R}(0)}{R(0)} \quad B^2 = \left[ \frac{\dot{R}(0)}{R(0)} \right]^2 - \frac{\ddot{R}(0)}{R(0)}
\]
The autocorrelation method

**However:** Accurate estimates of the derivatives of the autocorrelation function can be difficult to achieve. Therefore an alternative formulation is used:

Correlation function in polar form: \( R(\tau) = A(\tau) \exp[j\phi(\tau)] \)

Yields the following mean frequency and bandwidth estimate:

\[
\bar{\omega} = -j \frac{\dot{R}(0)}{R(0)} = \dot{\phi}(0)
\]

\[
\bar{\omega} \simeq \frac{\phi(T_{PRF}) - \phi(0)}{T_{PRF}} = \frac{1}{T_{PRF}} \arg[R(T_{PRF})]
\]

\[
B^2 = -\frac{\ddot{A}(0)}{A(0)} \approx \frac{2}{T_{PRF}^2} \left[ 1 - \frac{A(T_{PRF})}{A(0)} \right] = \frac{2}{T_{PRF}^2} \left[ 1 - \frac{|R(T_{PRF})|}{R(0)} \right]
\]

**In other words:** The power, mean frequency and bandwidth of the Doppler spectrum can be found using magnitude and phase estimates of the correlation function at lags 0 and 1 (T_{PRF})

Autocorrelation estimator properties

• Robust in low signal-to-noise environments
  – Superior to FFT-based method below ~15 dB, similar above ~15dB

• Computationally inexpensive
  – Ideally, in a noise free environment only two complex samples are needed to estimate the mean frequency
  – In practice more samples are needed to 1) attenuate clutter, and 2) reduce the variance of the correlation estimates
Cross-correlation method

- The velocity is proportional to the RF time shift between successive pulses

\[ \tau = \frac{2 \Delta z}{c} = \frac{2v \cos(\theta)T_{PRI}}{c} \]

\[ \hat{R}_{12}(m, m_0) = \frac{1}{N_S} \sum_{k=0}^{N_S-1} r_1(m_0 + k)r_2(m_0 + k + m) \]

\[ \hat{\tau}_{\text{max}} = \arg \max \hat{R}_{12}(m) / F_s \]

\[ \hat{v}_z = \frac{c \hat{\tau}_{\text{max}}}{2 T_{PRI}} \]

Auto correlation vs. Cross correlation method

Example:
*In vivo* Comparison of autocorrelation and cross-correlation for data from the human subclavian artery

The two methods are approximatively equal for narrow-band pulses and with radial averaging

Reference: Torp & al: Ultrasonic Symp. 93
Color mapping types

**Power Doppler (Angio)**
Brightness ~ mean power

**Color flow**
Brightness ~ mean power
Hue ~ Velocity

**Color flow ”Variance map”**
Hue & Brightness ~ Velocity
Green ~ velocity spread

*Image example:* Aortic regurgitation
Tissue / flow arbitration

- B-mode and CFI image acquired separately due to different resolution and penetration requirements in CFI and B-mode

- The two images are typically combined through a hard decision of whether to display a B-mode or color pixel \(\rightarrow\) Arbitration

- Post-processing is also needed in order to reduce the amount of flashing artifacts due to insufficient clutter attenuation

This decision is typically based on:
- Mean frequency and power before and after filtering
- High power / low frequency (below filter cut-off) \(\Rightarrow\) tissue
Tissue / flow arbitration

- B-mode and color-Doppler images are acquired separately due to different sensitivity requirements.
- The two images are combined through a hard decision of whether to display a B-mode or color pixel → *Arbitration*

**Power-Doppler:**
- **Hard arbitration**
- **Soft arbitration**

**Example:** Thyroid nodule vascularization
B-Flow imaging

- A power-Doppler variant
- B-mode and flow map from same data set, *no arbitration*
- Coded-excitation techniques to avoid loosing sensitivity
- *No color blooming*

Internal carotid stenosis  Common carotid ulceration  Dialysis Graft Pseudoaneurysms
Contrast enhanced ultrasound

- Microbubbles with a radius of 1 to 5 μm
- Stay in the blood pool
- Superior tissue signal separation
- Also suitable for perfusion imaging
Limitations of Doppler methods
Maximum measurable velocity

- The maximum measurable velocity is given by the temporal sampling frequency, i.e. the pulse repetition frequency (PRF), and the center frequency \((f_0)\), as:

\[
V_{\text{max}} = \frac{c \cdot \text{PRF}}{4f_0}
\]

- Cardiac imaging example:
  - PRF = 5kHz, \(f_0=2.5\text{MHz}\) → \(v_{\text{max}} = 0.77\text{m/s}\)
  - Covers normal flow velocities, but not high velocity jets

- Vascular imaging example:
  - PRF = 12kHz, \(f_0=5\text{MHz}\) → \(0.92\text{ m/s}\)
  - Still not covering high velocity jets due to stenosis
Minimum measurable velocity

- The *minimum* measurable velocity is *in theory* given by the requirement that at least one period of the carrier frequency needs to be observed:

\[
V_{\text{min}} = \frac{c \cdot \text{PRF}}{8 \cdot N \cdot f_0 \cdot \cos \theta}
\]

- However, *in practise* the cut-off frequency of the clutter filter will determine the minimum velocity
Aliasing and angle-dependency

- **Aliasing** is a sampling phenomenon that occurs when the velocity exceeds the maximum value:

\[ v_{\text{max}} = \frac{c \cdot \text{PRF}}{4f_0} \]

- **Angle-dependencies** occur as only the axial velocity component is measured.
What is aliasing

• Aliasing is a wrap-around effect that occurs due to finite sampling rate relative to the real movement

Sinusoid with two different frequencies from the same samples
Range-velocity limit

• The pulse repetition frequency (PRF) determines the maximum measurable velocity, but also the maximum imaging depth

• This imposes a range-velocity limit: \( z_{\text{max}} v_{\text{max}} = \frac{c^2}{8f_0 \cos \theta} \)

• Cardiac imaging example:
  – PRF = 5kHz, f0=2.5MHz, vmax = 0.77m/s, theta=0 \( \Rightarrow z_{\text{max}} = 11.8\text{cm} \)
  – Imaging of the heart can go deeper than 20 cm…
Range ambiguity

- It is possible to fire with a higher PRF than given by the range-velocity constraints

- If the PRF is higher than $c/2z$, then several range gates within the imaging field will be interrogated

- However, as long as the (high) velocity components of interest stands out in the spectrum, it can be of clinical value
Angle dependency in spectral Doppler

- The Doppler shift is dependent on the angle between the beam and flow
- Angle-correction can be done manually, by indicating the direction of the vessel flow
  - Works to sufficient accuracy as long as the angle of the beam is less than 60 deg compared to the flow
Flashing artifacts

• Flashing artifacts are false coloring of tissue regions

• Due to insufficient separation (filtering) of the blood

• In general a problem when the velocity of tissue and flow become comparable
  – Physiological movements
  – Probe movement during imaging

Example: Thyroid imaging
Other limitations

- Sensitivity vs. spatial resolution
  - Separate acquisition needed for flow imaging
  - Leads to **color blooming artifacts** where color covers the tissue

- Frame rate vs. image quality
  - Frame rate requirements reduce the image quality, leads to more smoothed images

- Accuracy of velocity measurements
  - A high variance and velocity dependent bias

=> **CDI is mainly used qualitatively**

Example: Carotid imaging using a broad image sector
Patient safety in Doppler imaging
Patient safety in Doppler imaging

- Potential hazardous heating and mechanical effects restrict the allowed acoustic output of ultrasonic imaging equipment.

- Acoustic output is restricted by one of the following:
  - Mechanical index (MI), a measure of mechanical effects (cavitation)
  - Spatial peak temporal average intensity (Ispta), output power, or thermal index (TI)
  - Transducer surface temperature

- Heating effects are averaged for combinations of modes. Mechanical index is determined by the modality with the highest value.

**The outcome:** The sensitivity / penetration in Doppler modes may be severely punished from these restrictions.
Mechanical effects

- Mechanical effects are related to cavitation, i.e. the formation and collapse of gas bubbles

**Mechanical index:**

\[
MI = \frac{p_{r.3}(z_{sp})}{\sqrt{f_{awf}}} \quad \text{Derated peak rarefractional pressure}
\]

\[
MI \quad \text{Acoustic working frequency}
\]

**Focal gain:**

\[
G_{\text{foc}} = \frac{D_a D_e}{R_{\text{foc}} \lambda} = \frac{D_e}{F_\# \lambda} = \frac{D_e \cdot f_0}{F_\# \cdot c}
\]

(Coinciding az. and el. Foci)

→ MI is measured at the axial point of maximum *derated* temporal averaged intensity \((I_{ta})\)

→ MI is effectively proportional to applied voltage, \(\sqrt{f_0}\), and inverse proportional to the F-number
Tissue heating

• Heating occur in the tissue due to absorption, at a rate much lower (∼seconds) than the ultrasound frame rate

• **Spatial-peak temporal average intensity (Ispta):**

\[
I_{\text{spta}} = \max_z \left[ \text{PRR} \cdot \int_0^{T_{\text{PRR}}} \frac{p_z^2(t)}{\rho c} \, dt \right]
\]

• Proportional to the transmitted pulse energy
• Proportional to the squared focal gain, \(G_{\text{foc}}^2\)
• For scanned modes, the intensity contributions for overlapping beams are averaged
Transducer surface heating

• **FDA regulations:** The transducer surface may not exceed 43 deg. held towards the skin, or 50 deg. towards air

• For current transducer technology, surface heating is almost always higher than at the geometric focus\(^1\)
  \[ \text{It is typically the most limiting factor with regards to voltage / PRF / apertures} \]

• The temperature rise for duplex and triplex modes (B-mode+CFI+PW-Doppler) is an average over a common time constant (frame rate / PRF)

PW-Doppler acoustics

- For a nonscanning situation, the maximum Ispta is close to the focal region
- The longer pulses typically used in PW-Doppler significantly increases heating effects
- For duplex operation (B-mode + PW-Doppler), surface temperature increase is mostly dominating
- For triplex operation (B-mode+CFI+PW-Doppler), Ispta or surface temperature will dominate

**Outcome:** PW-Doppler is typically Ispta limited, exceptions include very low PRFs, low frequencies, and narrow focusing

**Setup:** Linear array, f0 = 5MHz, FNtx = 2.5, txFocus 4 cm, PRF = 8kHz
Color-Doppler Imaging acoustics

- For a scanning situation, the maximum $Ispta$ is close to the surface.
- Similar characteristics for phased-array (cardiac) probes as for linear array (vascular) probes.
- Duplex operation adds heating effects from B-mode.
- Generally surface temperature limited.
- MI limited only for short pulses, low frequencies and PRFs, narrow focusing.

CDI is typically transducer surface temperature limited. Exceptions include very low PRFs, low frequencies, and narrow focusing.

Setup: Linear array, $f_0 = 5$MHz, $FNtx = 2.5$, txFocus $4$ cm, $PRF = 4$kHz, $2$ cm ROI width, $50$ beams. **NB: B-mode values not added**
Surface temperature comparison

- An example of surface temperature prediction for PW vs. CDI for the same aperture / tx-focus
  - CDI generates a higher surface temperature than PW-Doppler due to:
    - Aperture overlap during scanning, i.e. individual elements are excited more often,
    - The imaging PRF is often higher in CDI than for PW-Doppler

**Common setup:** Linear array, $f_0 = 5$MHz, FNtx = 2.5, txFocus 4 cm

**CDI:** PRF_img = 16kHz, PRF=4kHz, Np=12, 2 cm ROI width, 50 beams.

**PW:** PRF_PW = 8kHz, Np=64

**NB:** B-mode values not added
Patient safety summary

- Transducer surface heating is currently the main limitation in CDI and B-mode+PW-Doppler,
  - $I_{spta}$ is the main limitation in CW- and PW-Doppler (without B-mode)
  - MI limitations might occur for low PRFs, short pulses, and narrow focusing

- For an equal PRF, the temporally average transmitted energy is distributed over a larger spatial region for scanned modes than for PW-Doppler:
  - Lowering overall tissue heating effects for scanned modes
  - Bringing the maximum temperature close to the transducer for scanned modes
  - Especially the case for 3-D imaging

- These rules of thumb apply to both linear array (vascular) as well as phased array (cardiac) imaging
Recent advances
Higher acquisition rates
Background: Frame rate vs. image quality

In Color-Doppler Imaging a **sufficient frame rate** is often achieved on the expense of **reduced image quality and / or reduced image width**

Example: Carotid artery bifurcation imaging using a broad image sector. Notice the temporal acquisition lag from right to left in image due to the low frame rate (6 fps).
Ultrasound image formation
Conventional vs. parallel receive beamforming

Conventional image formation:
An image is built line by line, one for each ultrasound transmission

Parallel receive beamforming:
$N$ image lines are generated per ultrasound transmission
*Frame rate increased by $N$ times*
Plane-wave imaging
What’s up with that?

- Plane-waves are *unfocused ultrasound beams*, and therefore *without curvature* over a large field of depth

- One transmitted plane-wave can illuminate a large spatial region so that a **high number** of parallel receive beams can be utilized

- Due to the flat curvature, **image artifacts** normally associated with parallel receive beamforming for curved beams are **avoided**
What is the difference?

An illustration of the frame rate difference for the same amount of image lines (fully sampled). Playback is at 20% of real-time.

Example:

Neonate heart with an atrioventricular septum defect (AVSD).

Parasternal image cross-section using a linear array transducer

Roughly equivalent to current high-end acquisition, using 2x parallel beams

Plane wave imaging, 16x parallel beams (both B-mode and CFI)
Example: Vascular imaging

Stenosed carotid bifurcation of elderly male subject, an example of both higher frame rate and quality (2x lateral sampling) using plane-waves

- Conventional CDI acquisition, 2xPRB
- 2 flow images per B-mode image
- NB: 2 B-mode foci

No. of beams = 82, frame rate = 15 Hz

- Plane-wave CDI acquisition, 16xPRB
- 2 flow images per B-mode image
- Conventional B-mode acquisition

No. of beams = 192, frame rate = 54 Hz
Angle-independent imaging
Aliasing-free and angle-independent Doppler imaging – why?

- General improvements
  - In general less interpretation of color images, i.e. increased diagnostic certainty
  - No need for angle-correction, i.e. more accurate quantitative measurements

- New clinical information?
  - Improved volume flow quantification
  - Improved detection of shunt flow
  - Detection of circulatory / oscillatory flow, flow vorticity
  - Wall shear rate / shear stress estimation
Blood speckle imaging

- The speckle pattern from blood flow signal is isolated by high-pass filtering the Doppler signal.
- The movement of this speckle pattern is correlated to the movement of blood.

**Figure:** B-mode image of carotis artery and corresponding filtered image showing the blood flow speckle pattern utilized in BFI.
Blood Flow Imaging (BFI)

*Angle independent flow imaging*

- An *angle-independent* blood pattern visualization has been developed, compatible with existing methods.

Regular Color Flow Imaging  
Speckle + Color Flow Imaging
Intraoperative blood flow imaging in neurosurgery

Combination of real-time ultrasound imaging and navigation based on preoperative MRI during neurosurgery
Intraoperative imaging in coronary bypass surgery

Figure 6.3: Imaging of a proximally snared anastomosis (standard case) using CFI and BFI respectively.


*Blood Flow Imaging - A new 2D ultrasound modality for enhanced intraoperative visualization of blood flow patterns in coronary anastomoses*
New approach: BFI combined with plane wave imaging

By combining BFI with plane wave imaging, the best of both CDI and BFI is achieved at high frame rates, no trade-offs
Speckle tracking of blood

- Tracking speckle pattern between subsequent frames, speckle displacement is given by best match

\[
\varepsilon(\alpha, \beta, n) = \sum_{i=1}^{l} \sum_{j=1}^{k} |X_0(i, j) - X_n(i + \alpha, j + \beta)|
\]

\[
V_n = \frac{\sqrt{(\alpha_m \Delta x)^2 + (\beta_m \Delta z)^2}}{nT}
\]

\[
\theta_n = \arctan \left( \frac{\alpha_m \Delta x}{\beta_m \Delta z} \right)
\]

Compound vector-Doppler

Using parallel acquisition, the compounded frames are acquired at a very high speed:

- no loss in frame rate
- Near instantaneous acquisition of flow conditions
2-D velocity vector imaging

**Speckle tracking**

Quantifying the blood speckle movement using image pattern matching

Utilizing plane wave image acquisition, the accuracy can be substantially increased

**Example:** Vector velocity imaging of cardiac flow (systole) in a newborn
Example: Bidirectional ventricular septum defect (VSD)
Bidirectional VSD movie

Example:

Imaging of a ventricular septum defect (VSD) in a newborn baby

The VSD in this case leads to bidirectional flow between the ventricles

Speckle tracking used to quantify the blood movement

Streamlines used to visualize the velocity vector field
Low flow imaging
What is the clinical impact low-flow Doppler imaging?

- **Neovascularization**
  - Tumor malignancy
  - Vulnerable plaque identification, *vaso vasorum imaging*

- **Musculoskeletal**
  - Rheumatology
  - Tendinosis

- **Perfusion imaging**

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Neovascularization in a atherosclerotic pig model, *contrast enhanced US*

Neovascularization in patella tendinosis
Minimum measurable velocity

- The **minimum measurable** velocity is **in theory** given by
  \[
  v_{\text{min}} = \frac{c \cdot \text{PRF}}{8 \cdot N \cdot f_0 \cdot \cos \theta}
  \]

- **Example:**
  - PRF=1kHz, N=12, f0=10MHz, \( \theta = 0 \), c=1540 m/s
  - \( v_{\text{min}} = 0.16 \text{ cm/s} \)

  However, in practise the wall filter will determine the minimum velocity that can be measured
High-frequency imaging

- Commercial systems and transducer technology are currently available which allow for imaging at very high frequencies and at a high frame rate.

Examples: Imaging of venous flow in a healthy volunteer using a 16 MHz linear array transducer prototype. At high frequencies blood scattering may be visible in the B-mode images.
High frequency imaging

- Very high-frequency array transducers are now available for animal imaging
  - **Examples:** Visualsonics Vevo 2100, 9 - 70 MHz array transducers available

**Figure:** Imaging of mouse carotid at 40MHz, vessel size < 1 mm

**Figure:** Imaging of a mouse kidney at 33 MHz
Plane wave advantage for low flow

*Improved clutter filtering*

- By increasing both the *packet size* and PRF we can:
  - Achieve a high Nyquist limit avoiding aliasing
  - Keep the low velocities (same filter limits as before)

![Clutter filter frequency response](image_url)

*Keep low velocities*

5x velocity range

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www.ntnu.no
Extended Nyquist - Thyroid imaging

Healthy thyroid
PRF = 4 kHz
Packet size = 12

Insufficient signal
No aliasing
No flashing
Real-time 3-D flow imaging
3D color flow imaging with 2-D matrix array probe

From 1D Array…

64 – 128 array elements

Azimuth

Elevation

to 2D Array…

2000 – 3000 array elements
Real-time 3-D CFI example

Example:
Mitral valve insufficiency
Real-time CFI without ECG triggering and volume stitching
Main challenge is frame rate
Real-time 3-D CFI example

Example:

Mitral valve insufficiency

Real-time CFI without ECG triggering and volume stitching

6-slice view – notice the banana shaped leakage in the mitral valve indicated by the green color (increased Doppler bandwidth)

New information!
Estimation of Valvular regurgitation Area by 3D HPRF Doppler

Valvular regurgitant jet area can be quantified by isolating the Doppler signal power from the jet over the whole leakage area relative to a known reference signal from the center of the jet (vena contracta)
In vivo - compared to MRI

- Vena contracta position determined with PW Doppler.
- Jet power isolated with high pulse repetition frequency (HPRF).
- Acquire 3D HPRF data.
- Obtains jet geometry, cross sectional area, and regurgitant volume.
- Small leakages overestimated when compared to MRI. This is expected due to the finite beam width.

Regurgitant volume (RV) = area * velocity-time-integral

T. Hergum, T. Skaug
Emerging ultrasound transducer technology

- Higher frequency 2-D matrix phased-array transducers are currently on the market
  - Phillips X7-2 (2-7 MHz)

- 2-D matrix linear array transducers will be available in the coming years
  - High frequency 3-D imaging
Miniaturization: From high-end to hand-held

A dramatic reduction in the scanner footprint has become feasible through miniaturization of the electronic components.
Thank you for your attention

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