Signal Acquisition and Processing in Medical Diagnostic Ultrasound

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The use of ultrasound for medical imaging has been studied since the early 1950s. During this time, advances in technology and clinical practice have made ultrasound a leading medical diagnostic imaging modality. Modern ultrasound equipment is relatively inexpensive and portable, and it provides high-resolution real-time imaging without the use of ionizing radiation. The cost effectiveness and portability of the modality is particularly important in developing countries (and rural areas of developed countries) as it allows widespread access to sophisticated medical imaging in a timely manner. Today, ultrasonic imaging is used in a variety of clinical applications, including cardiology, obstetrics and gynecology, general abdominal imaging, and vascular imaging. Though ultrasound is generally regarded as a noninvasive imaging technology, it is finding increased use in surgical and intravascular applications, and in guiding other interventional procedures.

As uses for ultrasound increase, so does the need for effective signal processing techniques. In this article we provide an overview of signal acquisition and processing in modern medical ultrasound imaging equipment and present some of the more important concepts in practice today. Images and data types typical of current ultrasound equipment are reviewed; fundamental acoustics and design parameters that affect signal quality and processing strategies are discussed; and current research directions are presented.

Ultrasound Basics

Essentially all ultrasound scanners in clinical use today rely on echo imaging, as opposed to transmission imaging or some other method. An acoustic wave is launched into the body using a handheld transducer, the wave interacts with tissue and blood, and some of the transmitted energy returns to the transducer to be detected by the instrument. If we know the velocity of propagation in the tissue being interrogated, we can determine the distance from the transducer at which the interaction occurred. The characteristics of the return signal (amplitude, phase, etc.) provide information on the nature of the interaction, and hence they give some indication of the type of medium in which they occurred. This is the basis of ultrasonic imaging.

Ultrasonic imaging is, of course, concerned with making pictures. Figures 1 through 4 show several representative images illustrating what one might see being produced on modern equipment. Figure 1 shows a 4-chamber apical view of a heart using only echo information. This sort of display is referred to as a "B-mode" or "brightness mode" display. The left ventricle appears at the top of the image, with the left atrium in the bottom right. The right ventricle and atrium appear on the left side of the image. In real-time imaging, one would see the heart contracting and the valves separating the ventricles and atria opening and closing. The acquisition frame rate in this case was 31 Hz, which is typical.

Another way of viewing echo information is to only interrogate tissue in one direction and to display the resulting one-dimensional (1D) data over time as a scrolling display.

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Such a display is referred to as an "M-mode" or "motion mode" display. Figure 2 shows a B-mode image of a short-axis view of the left ventricle of a heart, with an "M-line" showing the direction of interrogation for the scrolling display at the bottom of the image. The advantage of M-mode imaging is that it has much finer temporal resolution (on the order of 1000 lines per second) than B-mode imaging, which, for example, can be important when analyzing cardiac valve motion.

In addition to simple "brightness" displays of echo information, ultrasound systems can detect motion using either continuous-wave (CW) or pulsed-wave (PW) Doppler techniques. The resulting data are usually processed and displayed as either spectral Doppler (Fourier transform information), or color flow images. Figure 3 shows both types of displays. The upper part of the image shows a zoomed B-mode image of blood vessels in the kidney, with overlaid color indicating blood flow. Traditionally, shades of blue are used for flow away from the transducer, and shades of red are used for flow toward the transducer. The bottom part of the image shows a scrolling spectral Doppler display. Here the height of the display indicates Doppler shift (or velocity), and the brightness of the display indicates the strength of the signal at a particular velocity.

Power-mode Doppler techniques are related to color flow imaging, but in this case an estimate is made of the power of the Doppler signal, as opposed to velocity and direction. This is an easier estimate to make and is generally more sensitive to low flow states and less angle dependent than color flow. These advantages allow more complete visualization of fine vascular architecture. Figure 4 shows a healthy adult kidney using power-mode imaging.

Fundamental Acoustics

The study of medical diagnostic ultrasound begins with an understanding of the physical principles involved. Most of medical ultrasonic imaging is well described through the analysis of compressional waves and their interaction with tissue and blood. Such waves may be easily excited and launched into tissue via an ultrasonic transducer, after which the energy in the wave is absorbed, reflected, or scattered by the tissue. Some of the energy that is reflected or scattered may then be detected via an ultrasonic transducer (most commonly the same one that launched the interrogating wave). The received data represent the interaction of the acoustic wave with biological media. To understand how to form an image, we need to understand the speed of propagation in tissue, attenuation of the acoustic waves, and what the return signals actually represent.

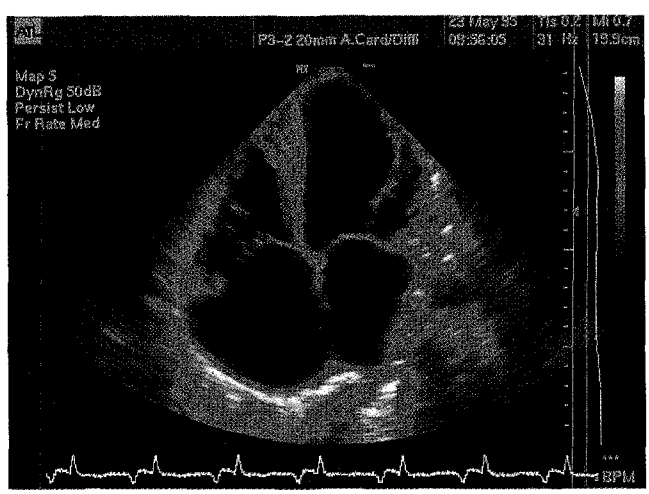
Compressional waves propagate through tissue with some characteristic velocity. The speed of sound in tissue varies with tissue type, temperature, and pressure. We can usually assume normal body temperature and pressure, so we are most concerned about tissue type. Table 1 gives some examples of acoustic properties of various media [1, pp. 5-25] [2, p. 552]. The values in Table 1 are representative and are reported with some variation in the literature. Mean sound velocity in human tissue is usually taken to be 1540 meters per second [1, p. 6].

Because some part of the energy of the interrogating wave is absorbed, scattered, or reflected continuously as the wave travels through tissue, the wave is attenuated as it travels deeper into the body. Attenuation occurs through a variety of mechanisms, but primarily due to absorption and scattering [3]. Attenuation is an exponential function of distance, often modeled as $A(x) = A_0\, e^{-\alpha x}$ where A is amplitude, A_0 is a constant, α is the attenuation coefficient in Nepers (which depends on frequency), and x is distance [4]. There is a component of attenuation that does not depend on frequency, but this is usually negligible. This model of attenuation has some fairly severe implications at general-purpose imaging frequencies (2 to 10 MHz). Tissue has an attenuation of roughly 0.75 dB/cm/MHz one-way, so the intensity halves every 0.8 cm at 5 MHz.

Acoustic energy returns to the transducer through two main mechanisms: *specular reflection* and *scattering*. Specular reflection is due to changes in *acoustic impedance* across interfaces that are significantly larger in extent than the acoustic wavelength [2, pp. 300ff]. Specific acoustic impedance, usually just referred to as acoustic impedance, has units of pressure per unit velocity. Just as with an electromagnetic wave in a transmission line, when an acoustic wave moves from a medium with impedance Z_1 to a medium with impedance Z_2, part of the wave will be reflected, and part of the wave will be transmitted. As with a transmission line, the reflection coefficient is

$$\Gamma = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

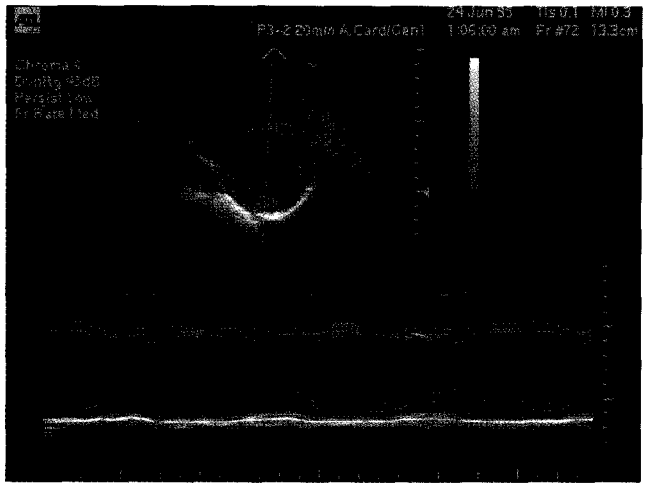
If the incident wave is not normal to the interface between the two media, the transmitted wave will be refracted accord-



1. A "B-mode" ultrasound image showing an apical 4-chamber view of a heart. The dark area that is nearly centered at the top of the image is the blood pool of the left ventricle, the right ventricle appears to the left of it in this presentation, and the atria are seen below. The valve leaflets separating the atria and ventricles are clearly shown.

ing to Snell's law. Note that the magnitude of the signal received by the transducer from a specular reflector will be orientation dependent. A structure may return a strong signal when insonified from a direction normal to its surface and may return a very weak signal if interrogated from a nearly perpendicular orientation.

Scattering occurs when acoustic waves interact with structures of size comparable to or less than the acoustic wavelength [2, pp. 300ff]. When insonified, these structures will tend to reflect weak waves in all directions (Rayleigh scattering). A volume of scatterers (like blood cells or organ tissue) will then act like a diffuse reflector [5]. The magnitude of the returned signal from a scattering volume may or may not depend on orientation, as opposed to specular reflecting interfaces. For example, scattering from muscle is orientation-dependent due to striation, whereas scattering from healthy liver tissue is not particularly orientation-dependent.



2. An "M-mode" display. The line in the B-mode image in the upper part of the display indicates where the data is being collected. In this case, the image shows the M-line bisecting a cross-section of a left ventricle. The lower display shows motion versus time; the "waviness" indicates cyclic motion as the heart beats. On this instrument, the lower display normally scrolls from right to left.

Ultrasonic Imaging

In the next few sections we will discuss the major functional blocks of a typical commercial ultrasound scanner, focusing on tradeoffs that affect signal characteristics and processing strategies. A simplified block diagram of a typical instrument appears in Fig. 5.

Pulser

The pulser provides the electrical waveforms that drive individual elements in the transducer. The intensity and spectral content of the acoustic signal launched from the transducer can be controlled by varying the center frequency, shape, duration, and amplitude of the pulser signal. General imaging equipment typically uses a center frequency of 2 to 10 MHz, though frequencies upwards of 40 MHz are used in special applications. Broadband pulses (i.e., short duration) are used for anatomical imaging, while narrowband pulses (up to and including CW) are more useful for flow imaging. Wave shapes may be as simple as monopolar square waves, or of arbitrary complexity. Signal amplitudes of 2 to nearly 200 volts are typically delivered to transducers with a nominal impedance of 50 ohms.

Transducers

The vast majority of transducers are made of one or more piezoelectric elements, with some sort of acoustic lens or protective material bonded to the front surface and a backing material bonded to the back surface. Piezoelectric ceramics are by far the most commonly used active materials. The active element(s) and any associated mechanical or electrical components are encased in a sealed plastic handle. Electrical connection is normally made via a bundle of microfine coaxial cables terminating in a low insertion force connector.

The simplest transducer is a *single-piston* transducer, which only has a single element. This element is usually circular and has some curvature for focusing of the acoustic wave. This element may be scanned mechanically to collect enough information to form an image, or held in one position if only 1D (e.g., flow) information is to be acquired. The primary advantage here is simplicity, with disadvantages including fixed focus, the need for mechanical scanning, and the difficulty in acquiring Doppler information from a moving transducer. Single-piston transducers are an older technology and are not normally used for imaging today. They are still found in "static" Doppler probes, which are used in certain adult cardiac examinations and in fetal heart rate monitoring applications.

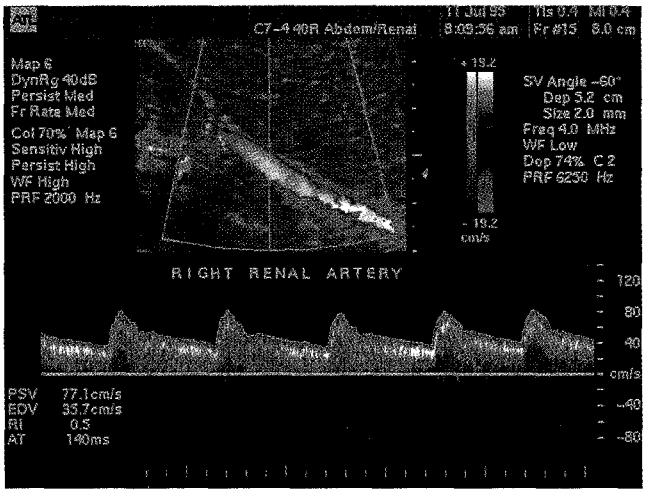
An improvement on the single-piston transducer is the *annular array*. This type of transducer is made up of several concentric piezoelectric rings, again with some curvature for focus. Such a transducer may be focused electronically on transmit and receive by offsetting transmit pulses to the rings, and delay-and-sum beamforming signals from the rings on receive. Again, the array is scanned mechanically to form an image. Annular arrays offer some of the best *elevational focus* (i.e., slice thickness) available. The requirement for mechanical scanning is still a disadvantage.

The dominant scanhead technology today is solid-state array technology. In these probes, a large number (usually between 48 and 200) of active elements are used to transmit and receive steered and focused acoustic beams. A *phased array* normally has a small (e.g., 15 mm) aperture, all of which is used to form beams emanating from one point. Phased arrays are normally used in cardiology applications (because the small aperture allows access between the ribs), and in deep abdominal exams. A *linear array* typically has a larger aperture (typically around 40 mm, but can be much larger) and more elements, which are used to form beams normal to the (flat) surface of the transducer. These transducers are used in a variety of abdominal, peripheral,

and small-parts exams. A *curved linear array* is a linear array formed against a convex curved face, giving a wider field of view since the beams formed normal to the surface of the array now spread out.

Beamforming

The primary advantage of multi-element arrays is the ability to electronically focus and/or steer the acoustic beam. The term "beamforming" is usually applied to focusing of the received data, though focus on transmit is also important. Normally, one transmit focal point is chosen for a burst, and continuous (or at least relatively continuous) focusing is performed on receive. A way to get more than one transmit focal point is to use a time-multiplex, or "multizone" technique. Several transmit bursts are sent, with data from depths in the vicinity of the previous burst focal zone being collected between bursts. If there is not enough "dead time" between successive bursts in a multizone image, range ambiguity artifacts may appear.



3. Color flow and spectral Doppler. The upper image shows a B-mode rendition of kidney structure, with overlaid color indicating blood flow in the renal artery. The color scale to the right of the image allows estimation of flow velocity. The white line in the center of the 2D image shows the placement of a Doppler sample volume in the artery. The lower display is a spectrograph of Doppler data collected within the sample volume. Such a display may be calibrated in frequency or velocity. On this instrument, the lower display normally scrolls from right to left.

Table 1. Acoustic properties of selected media

Medium	Speed of Sound (m/Sec)	Impedance 10^6 kg/m^2-S	Attenuation dB/cm at 1 MHz
Air	344	0.0004	12.0
Water	1480	1.48	.0025
Fat	1410	1.38	0.63
Muscle	1566	1.70	1.3-3.3
Liver	1540	1.65	0.94
Bone	4080	7.80	20.0

On receive, the incoming per-element signals usually pass through a time gain compensation (TGC) amplifier. The purpose of these amplifiers is primarily to compensate for depth-dependent attenuation of the signals. TGC is usually controlled by the operator with a set of slide-pots. Receive beamforming in commercial equipment is typically of the delay-and-sum variety [6, pp. 112ff.], with some variations such as phase-delay vs. true time delay [7]. Other techniques, such as frequency-domain beamforming, have not received wide acceptance. A distinction that is important in the industry is whether a beamformer is digital, hybrid, or analog. A digital beamformer will have analog-to-digital converters immediately following the TGC amplifiers, and all focusing delays are implemented digitally. A hybrid beamformer may introduce analog mixers for fine delay and/or basebanding. An analog beamformer uses analog delay elements for focusing. Digital beamformers have advantages in acquisition flexibility and broad bandwidth, while analog beamformers are still somewhat cheaper to produce.

No beamforming technique is perfect; the acoustic beam always has some nonzero depth-dependent extent in the azimuthal and elevational directions. This means that a particular pixel in the image represents information from some spatially weighted volume, as opposed to a point in space. Contributions from reflectors outside the main lobe of the acoustic beam thus give rise to *sidelobe artifacts*. Main lobe width determines the lateral resolution of the instrument, and the sidelobe level determines the clutter-free dynamic range of the instrument.

Array imaging relies on coherent illumination and detection. Just as with a laser, this results in interference artifacts known loosely as *speckle* [8, 9]. Speckle manifests itself as a random mottling of the image with bright and dark spots, which obscures fine detail and degrades the detectability of low-contrast lesions. It is unclear whether there is clinically relevant information in speckle patterns themselves; many experienced clinicians are very opposed to speckle-reduction techniques. Many different methods have been proposed for speckle reduction, such as adaptive spatial filtering, or combining multiple images with spatial or frequency diversity. Some speckle-reduction techniques have been shown to improve clinical image quality, particularly in terms of low-contrast lesion detection [10, 11].

Signal Processing

The composite beamformed signal is subjected to a variety of signal processing operations before it is presented to the display subsystem. Most modern equipment implements these functions digitally. We will consider the processing of three main data types here: echo information, spectral Doppler information, and color flow information.

The fundamental problem in echo imaging is to take the composite beamformed signal (something like 16-bit samples at a 20 MHz sample rate) and produce a brightness image

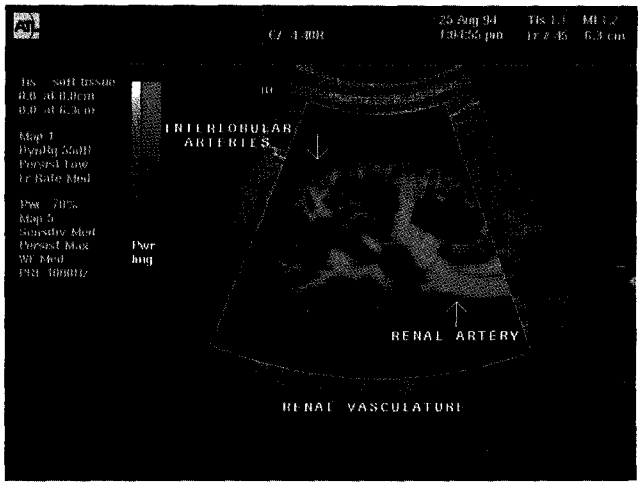
suitable for display on commercially available equipment. The composite signal is typically filtered, envelope detected, and then log-compressed to 8-bit resolution. Since the number of acquisition samples varies with the sample rate and imaging depth, some sort of multirate filtering is employed to match the display resolution to the acoustic data, and thus avoid spatial aliasing. The precise order of operations, filter design, etc., determines the quality of the final image, and thus is closely held proprietary information of the manufacturers, but the scheme outlined above will produce a reasonable image.

Two main acquisition techniques are used to produce spectral Doppler displays: CW acquisition and PW acquisition. In a CW system, a wave of a particular frequency is transmitted and simultaneously received either on a separated transducer or (usually) on a part of the transducer aperture that is not being used for transmit in an electronic array. The return signal is demodulated with respect to the transmit frequency, filtered, Fourier transformed, and formatted for display. The details of filtering, windowing, transform length, etc., are proprietary in commercial equipment, but good analyses are presented in chapter 9 of [12] and chapter 5 of [13]. A similar approach is taken in PW processing, though it is important to realize that in this case a position shift is being measured, rather than a true Doppler frequency shift. The details of this analysis are covered thoroughly in chapter 6 of [13].

Most instruments that have spectral Doppler capability provide an audio representation of the flow signal in addition to the spectral display. When demodulated, the Doppler signals from human blood flow fall into the audio range. While these signals can be used directly, most equipment separates forward and reverse flow into two channels and presents the information as a stereo audio signal. The separation into forward and reverse channels may be accomplished through a Hilbert transform technique described in [12, p. 96].

Because spectral Doppler information is collected at a relatively low data rate, we have the luxury of spending quite a bit of time processing it for display. Color flow data, however, typically has a much higher data rate due to the large number of sample volumes being interrogated, and we are unable to collect enough data for meaningful Fourier analysis at each sample volume. Instead, one typically uses a correlation technique for spectral analysis. Several acquisition lines (usually 3 to 12) are fired in the same direction, and the phase shift information at each resolution cell is used to estimate direction and average velocity. This technique was first proposed in [14]. Estimates of power and variance may also be computed; in fact, power-mode Doppler techniques only display the power estimate. An alternative technique proposed in [15] uses one- and two-dimensional (2D) time-shift estimation.

In all Doppler modes, differentiation of blood flow signals from tissue motion signals is important. Tissue motion signals are generally large in amplitude and low in Doppler frequency when compared to blood flow signals. Such signals result in low frequency spikes (and audio "thumps") in spectral displays and "flash" artifacts in color flow systems. A simple high-pass filter often gives reasonable results, but arbitrarily sophisticated adaptive filters may be employed.



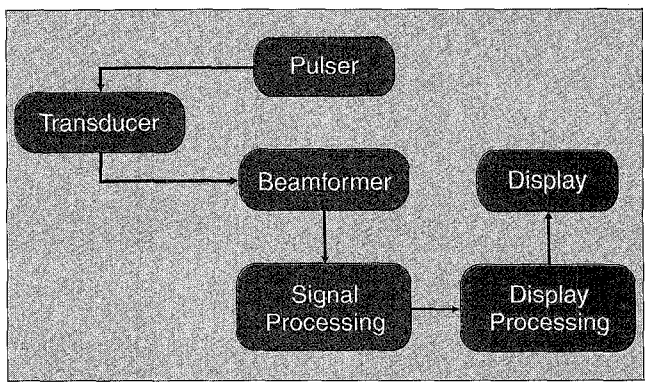
4. A power-mode Doppler display. This image shows vascular structure in a healthy kidney. This is similar to a color flow display, but higher sensitivity allows visualization of fine vascular structure. Notice that the color scale has no velocity annotation and that there is no directional information provided.

Display Processing

The function of the display block is to provide geometric correction, color and gray-scale mapping, and other display formatting services. Acoustic data are almost never collected at the same density as output display pixels. At a minimum, some sort of interpolation is needed to rescale the data for display. Except in the case of the linear array, data are collected in a polar coordinate system, requiring a coordinate transform and interpolation for display. It is interesting to note that interpolation and geometric correction can occur at many places in the signal chain, though most commercial equipment implements it after the signal processing block. Scaling and geometric conversion are generally referred to as *scan conversion*. Most commercial machines also allow the user to select from a variety of gray-scale and color-scale maps for display.

Current Directions in Research

There is a large ultrasound research community working on a variety of problems, and this review does not attempt to be all-inclusive in listing interesting and relevant topics. Rather, we wish to point out a few interesting areas for further study. Before launching into ultrasound research, however, it is important to keep in mind that the ultimate goal is to provide the clinician with the best possible information with which to make a diagnosis. There are varying opinions on what makes an "excellent" quality image, but there is agreement on some general principles. The following list of desirable features is



5. A simplified block diagram of a typical medical ultrasound scanner. The transducer converts electrical signals produced by the pulser into acoustic energy and converts returning acoustic energy into electrical signals. Many such signals are used in the beamforming process, and the resulting composite signal is processed and formatted for display.

an abbreviated and rearranged version of that given by Powis and Powis [16, pp. 393ff.].

- Like tissues look alike, and unlike tissues look different.
- Cystic structures look cystic and solid structures look solid.
- Low attenuators show acoustic enhancement and high attenuators show shadowing.
- Boundaries of organs and structures are visible.

These elements of image quality indicate the importance of understanding how a clinician is going to use an image. For example, one might decide that it would be a good idea to process an image to remove acoustic shadowing and enhancement behind cystic structures, though that is extremely valuable information for the clinician. A speckle-reduction algorithm might obscure subtle textural differences in tissue that may indicate pathology. Spatially variant contrast-enhancement procedures may make identical tissue types look different. The message is that one should consult the prospective clinical audience before "improving" the image formation process.

Beamforming

Biological tissue is not homogeneous, and therefore there will be variations in the speed of sound as an acoustic wave travels through the tissue. In addition, waves from individual transducer array elements propagate via slightly different paths to a target, and thus have different delay characteristics introduced by the nonhomogenaity. This causes imprecision in the focusing process. Correcting these errors is called *aberration correction*. This is a difficult problem that has not been solved satisfactorily to date. Some representative literature includes [17-21].

Medical ultrasound today is typically 2D—that is, a 2D B-mode or flow image is the primary means of visualization, and hence acquisition systems are geared toward this end. Real-time 3D image acquisition is problematic, as the speed of sound limits the number of transmit/receive intervals that can be squeezed into a reasonable frame time. Parallel formation of multiple receive beams from a single transmit burst is an approach that has been widely investigated. Representative literature includes [22] and [23].

Echo Signal Processing

Robust methods for tissue characterization and identification are, in some sense, the "Holy Grail" of ultrasound research. For example, clinically acceptable techniques do not yet exist that allow automated differentiation of normal and cancerous tissue. The solution of this problem will undoubtedly require a solid understanding of both ultrasound physics, interaction with biological media, and signal processing techniques. As such, it is one of the most challenging and interesting areas of research. A good overview of the topic is presented in [24] and [25].

Contrast agents are in common use in most areas of medical imaging other than ultrasound. Such agents allow enhanced visualization of structure, flow, or physiological processes. Recent investigations into the use of microbubbles as ultrasound contrast agents have provided very encouraging results. These agents can be as simple as free bubbles in an agitated saline solution, but they usually have a shell of Albumin or sugar for stability. The use of such contrast agents requires a new set of signal processing algorithms to optimally detect their presence. For example, it has recently been shown that some agents exhibit harmonic behavior—that is, if insonified at a particular (characteristic) frequency, they will emit harmonics of that frequency that can be detected using conventional equipment. This allows precise localization of the agent, which can be very valuable in the investigation of blood perfusion. Again, this is an area that will require good knowledge of both the physics and signal processing of the problem in order to develop satisfactory solutions. Some recent literature includes [26-28].

Because ultrasonic images exhibit speckle artifacts, speckle-reduction techniques have been a popular area of research. Some of the more successful techniques have been *compounding* techniques, where two or more signals representing the same spatial area are computed and merged to form a composite image. *Frequency compounding* involves processing the composite beamformed signal over two or more separate frequency ranges (within the bandwidth of the transducer) and merging the results. *Spatial compounding* involves processing signals covering the same spatial area from at least two different look directions and merging the results. *Temporal compounding* (a.k.a. frame averaging) composites images over time. Note that this has no effect if there is no motion in the image. Other image processing techniques, such as the use of nonlinear rank-order filtering, have been investigated to some extent, without much success. Representative literature includes [29-35].

Doppler Signal Processing

Traditional Doppler techniques can only estimate motion normal to the insonifying field. Most vessels, however, are not in this orientation. Manual techniques for angle correction are found on all modern equipment, but what is really wanted is a technique that provides Doppler information as a vector rather than a scalar. The two techniques that have been most promising in this area are the use of two or more beams per sample volume and 2D correlation techniques. Some papers on multiple-beam methods include [36] and [37], while [38] and [39] discuss correlation techniques.

Another area of active research is improved spectral estimation techniques for color flow imaging. At this point, color flow provides an excellent spatial visualization of flow trends, but the precision and variance of the estimators is not good enough to be reliably quantitative in a clinical setting. More research is needed on methods that can provide true quantitative color flow at reasonable frame rates. For a start in this area, see the overview in chapters 7 and 8 of [13].

Advanced Visualization

The goal of any medical imaging technique is to convey clinical information effectively. While traditional ultrasound image displays are extremely valuable, there has been increasing interest in new methods for visualization of ultrasound data. There has been particular interest in visualizing the spatial relationships between successively acquired images, and the added clinical utility that such techniques offer. Commercial manufacturers have developed techniques for 2D spatial registration of images (using a spatial correlation technique), and 3D visualization using volume rendering techniques [40 - 41]. The 2D technique allows a wider field of view than would otherwise be possible, and 3D visualization shows promise in the evaluation of both structural and vascular anatomy.

Conclusion

The development of signal processing techniques for medical ultrasound is a challenging and rewarding pursuit. The development of effective techniques generally requires a good knowledge of ultrasound physics, clinical applications, and modern signal processing techniques. Ultrasound today has the best price-to-performance ratio of any medical imaging modality and thus will continue to receive added attention in cost-conscious health-care markets. In all, this promises a bright future for practical research in this field.

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