Resolution and Display Requirements for Ultrasound Doppler/Evaluation of the Heart in Children, Infants and the Unborn Human Fetus

DAVID J. SAHN, MD, FACC

La Jolla, California

Technical considerations and the instrumentation used for pediatric two-dimensional echocardiography and Doppler examination are reviewed. The configurations of sector scanners, the function of the mechanical versus phased array systems and considerations related to lateral, axial and azimuthal resolution requirements are discussed. The performance and requirements for echocardiographic scan converters and the requirements for pediatric display are reviewed. Methods of performing quantitative Doppler echocardiography are discussed because this technique provides new and important types of information for the evaluation of congenital heart disease. Considerations of Doppler velocity, Doppler spatial resolution and Doppler display requirements are presented. Characteristics of ultrasonic imaging devices for use in fetal echocardiography and fetal Doppler study are reviewed, and a brief overview of techniques for the extraction of information about the nature of ultrasound scatterers (that is, tissue signature) is presented. It is the purpose of this technically oriented discussion to present the capabilities, trade-offs and needs for future development relevant to pediatric echocardiography in 1983.

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Configuration of Ultrasound Scanners for Pediatric Echocardiography

The initial real time ultrasound scanners were large aperture linear arrays that provided capabilities for real time two-dimensional imaging (1). These suffered severely, however, from lack of maneuverability and rib artifacts (distortion, reverberation and shadowing). Cardiac scanning devices rapidly moved toward a sector scan configuration, which allowed improved access to intercostal spaces with examinations performed very much like previous M-mode sweeps. New windows became available, however, because the two-dimensional format allowed an understanding of anatomy viewed from the apical, subcostal and suprasternal windows. Three general types of ultrasound scanner configurations are currently in use for echocardiography. The first type represents mechanical scanners, the second type consists of phased array scanners and the third type includes a potential hybrid configuration under exploration by several manufacturers. Because of the highly competitive nature of the market, all these devices have been redesigned for smaller or easier to manipulate transducer configurations and higher resolution.

Mechanical sector scanners. Two basic mechanical scan configurations continue to be used. Most transducers are “in-line” “end fire” in type with small lightweight motor drives. In the first type, the transducer is oscillated back
display of lines of sight between the forward and backward sweeps, which are each performed at 15 to 30 frames/s. Most of these transducers are linearly driven so that they are not slowed down before changing direction, and since about 1979 to 1980, all have been enclosed in short oil paths so that no vibration is felt by the patient. The oil path in these systems, however, has continued to provide ongoing problems with bubble formation, and precludes, for the most part, gas sterilization of these mechanical scanners. The second popular configuration for oil path-enclosed mechanical scanners is that of devices in which one, two or even three transducer elements rotate within the head in one direction all the time with, again, an interlaced raster, usually at 30 frames/s. Although these multitransducer devices have the potential capabilities of multifrequency interrogation, they have not yet achieved configurations as small as some of the rocking mechanical scan heads.

**Electronic sector scanners.** After the initial development of phased array sector scanners in Holland by Somer (2) and the evolution of the first cardiac phased array scanners by VonRamm and Thurstone (3), early difficulties were encountered because of the extreme electronic complexity of these devices, the rapid and accurate timing pulses needed for phasing and focusing and the difficulties in cutting and actually producing small high frequency transducer arrays. In recent years, however, the ongoing revolution in solid state electronics and advances in transducer technology and materials have made available phased array devices capable of producing image quality equivalent if not superior to images from mechanical scanners.

The functioning of phased array instrumentation in relation to steering and focusing of ultrasound pulses should be reviewed in any of the standard ultrasound texts (4). Briefly, by using the elements within an array in groups, Huygen’s principle of waveform behavior in optics (relevant to other waveform energies as well) allows ultrasound technology to utilize the same principles initially made available in phased array radar. If the excitation pulses to an array are altered by microsecond delays to achieve an in-phase waveform summation, this results effectively in ultrasound wave propagation at an angle to the normal of the array, and the beam in its transmit sequence can effectively be steered. A similar concept allows transmit focusing of beams. Further, by biasing the reception phase of the instrumentation in a complicated manner to test the pattern of returning waveforms for a radius of curvature relating to a specific depth or origin, the ultrasound receiver can be biased to amplify targets at specific depths. This allows a reception focus in zones of depth or the reception focus can be changed continually during the receive mode to allow focusing of echoes in both near and far zones — a concept called “dynamic focusing.” Although complex, these abilities of phased array instrumentation have now become electronically less expensive and available in smaller, more reliable instrumentation packages.

**Mechanical scanners with annular arrays.** The third instrumentation configuration being explored represents a hybrid of those just discussed and is best described as the mechanically steered annular array. If the oscillating or rotating transducer within an ultrasound device is itself an array based on concentric rings, then the individual piezoelectric sections or rings can be viewed as array elements capable of being used for focusing on transmit-receive, or dynamic focusing. The steering in these devices is achieved by mechanically oscillating the entire annular array, thereby avoiding some of the side-lobe patterns or the directivity (directional sensitivity receiver bias) that produces artifacts in many commercially available phased array devices. In the interest of brevity, I will not discuss the individual trade-offs among automatic versus various manual gain, reject and other signal optimization circuitry.

**Resolution Within an Ultrasound System**

Resolution, that is, the ultimate performance measure of an ultrasound system, may be discussed within the subdivisions of axial, lateral and azimuthal (or elevational) resolution (Fig. 1).

**Axial resolution.** Axial resolution refers to the ability of the device to resolve a structure from the one behind it. Axial resolution has been substantially improved in the most recent instrumentation by the move toward greater sensitivity achievable at a higher frequency. In many systems, imaging can be performed in adults at 5 MHz and pediatric examinations can be performed at 7, or even 10 MHz. Power output and, to some extent, the sensitivity of these transducers are functions of the number of pulses with which the

**Figure 1.** The three types of ultrasound resolution for a sector scan. The distance between A and C is resolved by axial resolution, the distance between A and B is resolved by lateral resolution and the distance between A and D is resolved by azimuthal (or elevational) resolution. (Modified from an illustration prepared by Dr. Joseph Kisslo.)

![Resolution Diagram](image-url)
transducer is excited. The longer the period of excitation and, therefore, the longer the transmit period, the less accurate the received axial resolution. A smaller number of pulses or shorter bursts of ultrasound allow individual targets, one behind the other, to be resolved more accurately and result in improved axial resolution. The number of pulses can now be controlled by the user in some systems. Improved receiver sensitivity has allowed high frequency imaging with shorter pulse bursts and improved axial resolution.

Dynamic range. Axial resolution is also a function of dynamic range, that is, the ability of the device to detect high intensity as well as faint echoes. Structures that lie perpendicular to the sound beam are better scatterers and reflectors of ultrasound. Most commercially available devices provide between 40 and 60 dB of dynamic range, that is, the difference in the sound intensity between the strongest and weakest echoes that can be distinguished from noise, resolved and displayed. Such dynamic range resolution allows imaging of fine scatterers and texture within the myocardium, but still lies beyond the capabilities of most ultrasound systems having display within the usual 32 to 64 shades of gray. To allocate the display capabilities of the device to a receiver with a wide dynamic range, a variety of dynamic compression schemes is required.

Most ultrasound instruments functioning at 3.5 to 5 MHz have axial resolution capabilities in the submillimeter range, with 0.1 to 0.2 mm determined as a function of the wavelength itself and, therefore, related to the frequency of interrogation and the number of pulses sent out. It should be pointed out that all discussion relating to the resolution within an ultrasound instrument is relevant not only to two-dimensional displays, but also to the derived M-mode images available because the ultrasound beam in an M-mode configuration being plotted as a function of time has the same advantages and limitations of resolution as those of the individual lines that are combined into a planar sector raster in two-dimensional echocardiography.

Lateral resolution. This refers to the ability of the ultrasound beam to resolve structures lying next to each other. It is usually measured as the difference in intensity between the echo reflected from a target in the center beam and the intensity of the beam as it is translated laterally to a point where the same target produces an echo decreased 6 dB from its in-line center beam strength (Fig. 2). This definition of ± 6 dB beam width refers to water tank performance of an ultrasound system and provides only a rough estimate of performance in a patient. In mechanical scanners using a cylindrical single transducer, the determination of lateral resolution in any one plane generally refers to a cylindrical beam. Lateral resolution measurement also relates, therefore, to elevational resolution or the thickness of the slice being scanned. For acoustical transducers focused only by lenses, the transition between near field and far field occurs at a point determined by the square of the radius divided by the wavelength of the transducer, after which the beam pattern then diverges in the far field. An acoustical focus can be superimposed on such single transducers to vary the maximal depth of focus before which the beam later diverges. For an electronically or dynamically focused annular array scanner or a dynamically focused phased array, the length of the optimal focus within the scan plane can be extended to improve lateral resolution over most of the depth of scanning.

Elevation or azimuthal focus. For a phased array device focused within the plane of imaging by electronic or dynamic received focus, the determination of the slice thickness, the thickness of the plane being scanned or elevational resolution is different and determined separately from lateral resolution. At present in most of these devices, an acoustic focus is applied and the transducer may be labeled medium or long focus, referring only to the depth of the acoustically imposed elevation or azimuthal focus. This maximal elevational focus range is no different than the short focus imposed on a single crystal transducer beyond which the beam begins then to diverge in the azimuthal domain, degrading resolution in the far field.

Lateral resolution performance. The water tank measurements of most 3.5 to 5 MHz beam profiles in ultrasound systems suggest a 6 dB beam resolution of ± 1.5 mm. This might suggest that axial resolution is about 10 times better than lateral resolution. In practice, however, since the eye can discriminate easily within 6 dB of dynamic range, the
lateral resolution (for example, determining how small a ventricular septal defect can be viewed in a side to side direction) is usually about 1 to 1.5 mm or ± 0.6 to 0.7 mm. Achieving an axial definition of two boundaries obviously aids in defining smaller and smaller structures. Nonetheless, the inherent resolution of the individual lines within an M-mode image or the lines of site within the sector scan are, in fact, influenced greatly by the techniques used to display the information.

Display of Ultrasound Information

Most ultrasound devices now use a solid state video display that converts the sector scan raster into a rectilinear raster, assigning each of the positions within the polar coordinates of the sector scan a locus on the video screen. This technique is called "scan conversion," involving two steps. The first is the calculation of the position of the target, which can be done in an X-Y raster or by assignment of an r-theta location, (that is, a radial distance along an angular coordinate) (Fig. 3). Next, the target is assigned an intensity value based on a sampled video intensity. Each of the loci within the video display is represented by picture elements or pixels. The average raster contains 256 pixels along each line for 256 lines, but some may have a pixel matrix as large as 512 × 512. Once the location of an echo-producing target has been determined and it has been assigned to a specific pixel along a certain line, it can then be assigned an intensity as an analog or a digital value. The intensity value varies depending on the number of "bits" of memory within the scan converter, which is merely a computer for assembling and remembering whole individual video frames (Fig. 4). A one bit capacity provides a black versus white discrimination, a two bit capacity allows two intermediate gray scale levels and, as shown in Figure 4, allows the assignment of various shades of gray. Most ultrasound devices have at least a six bit, 64 gray scale memory.

Spatial and temporal averaging. A variety of cosmet-
can either read intensity as primary analog video information, or the information can be retained and read as fully digitized information for implementation of a variety of computer processing algorithms. This further manipulation of ultrasound images aimed at texture analysis, as well as automatic edge definition, will be discussed at the end of this report. The user should be aware, however, that the scan-converted information can be accessed, directly read into a computer or digitized into the computer memory offline for further manipulation.

**Notation on Fetal Echocardiography and Relevant Instrumentation Requirements**

A brief note should be made regarding the requirements for achieving adequate imaging of fetal structures. Obviously, the fetal heart is smaller than the pediatric heart and yet it does not lie directly underneath the transducer, but is separated by a large standoff (that is, the mother's abdominal wall, placental structures and uterine contents). The task of fetal imaging, therefore, becomes the task of achieving high resolution at depth. Nonetheless, a large contact surface is available on the abdominal wall of the mother, which is less limited by intercostal spaces or windows. Sector scanners with high resolution remain very popular in obstetrics, especially for evaluating the lower uterus. The capability of using large aperture devices (such as large electronically focused linear arrays), combinations of technologies (such as large footprint annular array mechanical sector scanners) or large compounding arrays (to produce interlaced rasters of sector scans with high-lined density and high resolution) has many advantages for fetal imaging. The imaging constraints of the size of the window and the need for high resolution at depth provide the opportunity of applying new technologies (such as annular arrays or compounding phased arrays) to the echocardiographic study of the fetus.

Some of the window limitations involving echocardiograms obtained through the chest wall do provide problems in fetuses of later gestational age in whom shadowing by the rib and spine limits the obtainable fetal cardiac detail. Because of these shadowing problems during late gestation, most detailed fetal cardiac structural analyses are performed in the second trimester. The configurations of fetal echocardiographic devices can, therefore, be the same or sometimes substantially different from those used for routine echocardiography. The configurations are also much more likely to be based on the larger apertures windows and the higher resolution at depth requirements. Improved sensitivity in transducer materials and improved electronics have made obstetric examinations available at higher scan frequencies than those used during the initial development of fetal echocardiography.

**Doppler Echocardiography**

A brief discussion is pertinent regarding principles of Doppler instrumentation. It will be subdivided into spatial resolution of Doppler systems, velocity resolution of Doppler systems and Doppler displays.

All Doppler instruments work on the Doppler principle, which is the property of any moving structure to backscatter energy at a frequency different from the frequency transmitted. In cardiac or vascular applications, circulating red blood cells backscatter ultrasonic waves emitted from a piezoelectric transducer so that the returning sound energy differs from the transmitted frequency by a function of the vector of red cell motion along the direction of interrogation. The Doppler equation can be summarized by:

\[ \text{Doppler blood flow velocity} = \frac{F_1 - F_0 \times \text{velocity of sound in blood}}{2 F_0 \times \cos \theta}, \]

where the velocity of sound in blood is 1.540 cm/s, \( F_1 \) = the received frequency, \( F_0 \) = the transmitted frequency and \( \theta \) is present because sound energy travels to and from the transducer. Cosine \( \theta \) relates to the angle of incidence between the Doppler sampling direction and the direction of flow, so that if flow were perpendicular to the direction of sampling, no shift is received, the cosine of 90° being 0. In contrast, if the flow is parallel at 0° or 180° to the direction of interrogation, the maximal Doppler shift can be detected.

**Modes of Doppler Interrogation**

Three modes of Doppler interrogation are currently in vogue (Fig. 5). It is probably best to discuss these modes and their abilities to resolve Doppler velocities and then to discuss separately spatial resolution in Doppler echocardiography.

**Continuous Doppler mode.** Historically, the first mode of Doppler interrogation was the continuous Doppler mode. This mode requires separate transmit and receive elements so that one piezoelectric element is continually sending and the other continually receiving energy. It is easy to achieve a continuous Doppler mode with split transducers, but this mode has also been achieved with phased arrays, even with simultaneous imaging. As shown in Figure 5, continuous Doppler sound energy is processed for the Doppler frequency shift along the course of the beam and is limited only by its gradual attenuation in terms of the strength of signals received from far away structures. There is no selection of the depth of Doppler processing and, as such, there is no range or depth gating. Nonetheless, because the Doppler ultrasound is continually sampled for velocity shift at rates limited only by the electronic determination of the Doppler shift frequency \( (F_1 - F_0) \) either by spectral analysis, multiple filters or other techniques, the continuous wave Doppler technique is capable of resolving very high
Figure 5. Three modes of Doppler interrogation of a pulmonary artery signal. For pulsed Doppler, a 6 cm sample volume may be placed distal to the pulmonary valve as shown in the interrogation lines on the left (denoted by the arrow to the middle set of labels). Maximal pulse repetition frequency (Max PRF) available at 6 cm is near 13,500. At 3.5 MHz, this would be a velocity resolution of ± 143 cm/s. In the interrogation line on the right, continuous interrogation has been performed adding Doppler shifts along the line of the pulmonary outflow, proximal and distal to the valve. The maximal detectable velocity (Max Vel) would be limited only by the spectral analyzer. For the interrogation line in the middle (denoted by the arrow leading to the lowest set of labels [a high pulse repetition frequency system]), a gate has been set at a depth of 2 cm, effectively allowing the pulse repetition frequency to be tripled compared with the 6 cm sample volume in single pulse mode. The signal from 6 cm deep would be returning within the Doppler sampling time from two pulses before the most recent one.

velocities, even those arising from stenotic valves producing a flow velocity up to 8 to 9 meters/s.

Pulsed or range-gated Doppler mode. The second type of Doppler interrogation developed (responsible for much of the interest in cardiology because it allowed selective examination of different areas of the heart at different depths) was pulsed or range-gated Doppler echocardiography (5). In this concept, a temporal gate is established during the returning echo signals so that the returning information is analyzed for Doppler shift only during the selected period. This represents a gate in time which, because of the constant speed of travel of sound energy in tissue, is at a known distance from the transducer. The length of the range gate is usually adjusted by the operator for between 0.1 and 2 cm of depth of information. By increasing the range gate length, more structural depth or vascular space is included within the sample volume and the signal to noise ratio of the Doppler signal is improved unless extraneous wall or valve echoes are received. A variety of filtering algorithms can be used to remove some of the low frequency signals from walls and valves that may be within the sample volume.

A direct relation exists between the depth at which Doppler information is being sampled and the maximal pulse repetition frequency that can be used to sample it (that is, how often one can sample if one has to listen for a specific

period of time to receive information from the known depth). This relation is the so-called "range-velocity product" which is discussed and diagrammed in detail elsewhere (6). In essence, all three features of the Doppler equation enter into the range-velocity product, so that velocity resolution can be increased at any specific depth by decreasing the transmit frequency of interrogation, $F_o$ (which appears in the denominator of the equation) or by increasing the angle of incidence (since the cosine of the angle of incidence also appears in the denominator). Sampling must be performed at a frequency at least two times the frequency of any velocity shift in order to unambiguously resolve forward versus reverse (phases of a sine wave) flow information. The so-called Nyquist frequency is equal to half of the pulse repetition frequency of sampling, and represents the highest velocity shift that can be resolved unambiguously by a pulsed Doppler device. As will be pointed out in this report, abnormal flow in stenotic or regurgitant lesions reaches velocities of 3, 4, 5 or 6 meters/s or higher. Such velocities are often beyond the velocity resolution available to pulsed Doppler instrumentation. As an example of the average pulsed Doppler device, one commercial scanner functioning at 13,500 pulses/s in a single range gate mode at a range of sampling from 0 to 6 cm/depth can resolve ± 143 cm/s or 1.43 meters/s. This allows it to quantitatively resolve physiologic aortic, pulmonary artery and atrioventricular valve information, but not the very high velocities present in pathologic flow.

High pulse repetition—range-gated Doppler mode. The third type of Doppler interrogation utilizes the principle of range ambiguity to place a number of sample volumes at fixed intervals which are multiples of a baseline depth of interrogation. For instance, if one is interested in resolving high velocity flow occurring at 6 cm from the transducer, one could potentially set a range gate for a rapid pulse repetition frequency to listen only to a depth of 1.5 cm and solve all the information coming back within that range gate or the equivalent time period after transmission. The information coming back from 6 cm depth from three pulses before the last one sent would, in fact, be present within that shorter time gate, but would have traveled four times as far. The concept of high pulse repetition frequency or multiple range gate sampling, therefore, allows one to select specific areas of interrogation. However, there remains the inherent problem of not knowing which multiple of the shortest depth is the source of any particular velocity shift and the potential problem of noise from slow-moving targets that may be present within the first gate. This intermediate solution, however, appears capable of allowing a pulsed Doppler system without double transducer technology to resolve high velocity flow.

Doppler spatial resolution (lateral dimensions of sample volume). Having discussed range gate length (that is, axial resolution and velocity resolution), I will now review
the second type of spatial Doppler resolution, which involves the lateral dimensions of the Doppler sample volume. As discussed, the combination of two-dimensional scanners with M-mode capability allows one to sample for M-mode display information along any line within the sector, either simultaneously or after stopping the mechanical transducer. The same is true of Doppler sampling now that Doppler instruments are integrated with two-dimensional devices. This allows sampling for Doppler shift along any selected line of information. The lateral dimensions of a particular line of information for Doppler sampling are, in fact, slightly different than they are for the plotting of echographic targets. However, the same analogy for a test system is relevant.

As described by Walker et al. (7), the determination of the lateral beam width of a Doppler sample is essentially the lateral displacement of the transducer required to produce a decrease to half maximal intensity (6 dB) in the amplitude strength of the Doppler shift signal from a moving string target. The Doppler shift frequency, of course, remains unchanged because the velocity of the string target remains constant. Since the scatterers producing the Doppler shift in clinical studies are red blood cells that are not even imaged on clinical echocardiograms, ultrasound systems appear to be more sensitive to Doppler information than they are to imaging information. Doppler sample widths are, therefore, wider than echographic beam widths, often ±2 mm at a depth of 4 to 6 cm. This needs to be considered during fetal Doppler examinations because the information from both great arteries or both atrioventricular valves can sometimes be added together if both are included within the beam width of the Doppler device. The course of the Doppler sample volume width as a function of depth (that is, the lateral resolution) is subject to the same considerations as for lateral resolution in two-dimensional images (that is, having a focal point and far field configuration). The Doppler beam can also be dynamically focused or, if necessary to cover a large sampling area, defocused. Although simultaneous imaging and Doppler sampling can be achieved by split mechanical scanners or phased array devices, the maximal pulse repetition frequency rate available should still be applied to the Doppler mode to increase resolution velocity. The image may be “skeletal” in terms of line density, or it can be used as an update image for structure localization purposes.

Displays

A variety of Doppler information displays have been in vogue depending on technical capabilities. In general, it has been our experience that the most useful and efficient displays for conveying information include those with Fourier transform or other types of true spectral output (Chirp Z, for instance, an analog means of spectral analysis). Spectral output contains information as to the arithmetic mode of velocity shift, the shift most commonly present within the Doppler sample volume, and also information on the spectral width or dispersion as a function of “organized or disorganized flow” and on other velocities that are present in the signal. The allocation of density bins to the spectral display depends on the proportion of the scatterers that are moving at different velocities. This can often be changed (Fig. 6) to provide different statistical measures in the display and determinations of a statistical mean or maximal velocity can be superimposed on the spectral tracing.

The recommendations of the American Society of Echocardiography’s Doppler Standards and Nomenclature Committee for Doppler Display suggest that the following information should be shown within a Doppler image:

“...It is recommended that Doppler outputs be shown as spectra. That flow towards the transducer be shown as positive and upward; flow away from the transducer, downward and negative.

In addition, the baseline frequency of interrogation, a positional cursor showing where the information is coming from, the depth of interrogation, the Nyquist frequency and an estimate of sample volume size and angle of interrogation should, likewise, be shown on a page printout of a Doppler device.”

This allows all information necessary to deal with the Doppler output to be presented in a single image. Likewise, although one could perform the KHz to velocity (cm/s) transformation in the Doppler equation on a hand calculator given all this information, the Committee has suggested that the calibration for velocity on the Doppler displays be in centimeters or meters/s. The Committee also suggests that whether the velocity shown has or has not been automatically corrected for the angle of incidence displayed should be indicated. This type of Doppler output and an early standardization of terminology in this area should improve

Figure 6. This diagram shows how a spectral display builds up a gray scale pattern of the component velocities within a signal. The velocity most frequently present, 50 cm/s, would be the modal velocity and is shown by the darkest line in the spectral display. The other velocities contribute to the envelope or the spectral width.
communication and application of this important new ad­
junct to cardiac ultrasonography.

Extraction of Tissue Information

The last area I will discuss is extraction of tissue char­
acterization information from ultrasonic images, usually re­
lating to the derivation of information concerning myo­
cardial fibrosis, collagen content, calcium content or the
possibility of the detection of ischemia and scar. Fast image
processing and computer technologies, which can allow au­
tomatic identification of boundaries and density gradients
and analysis of the statistical distribution of different types
of reflectors within the tissue, have provided new capabil­
ities in these areas related to computer scanning of the dig­
itized ultrasound image information. A brief overview of
this area would cover the types of approaches to myocardial
characterization.

Analysis of backscatter radio frequency data. One
approach relates to the radio frequency or raw informa­tion
coming out of the machine and read on individual sector
lines before any further processing. This was the first and,
perhaps, the most fruitful (for in vitro studies) and complex
method for evaluating the content of an ultrasound signal.
The information is rectified so that all waveforms within
the analog information are shown as positive. A judgment
as to how much energy comes back is made with reference
as to a standardized strong (steel plate) reflector to evaluate
the integrated backscatter returning from ultrasound targets.
It has been a generally demonstrated phenomenon that early
during ischemia, there is an increase in integrated back­
scatter, which is, in essence, higher attenuation allowing
less energy to get deeper into the tissue because more re­
fection takes place. Other investigators have employed a
histographic analysis of this radio frequency data or have
examined the dependence of backscatter on the frequency
of interrogation in experimental situations, deriving a char­
acteristic backscatter versus frequency curve for normal and
ischemic myocardial variables.

Another approach to analysis of backscatter radio fre­
cquency data involves the autocorrelation technique for eval­
uation of the spacing of scattered energy in the frequency
domain as a function of potential changes of fiber distri­
bution. This was extensively explored by Mimbhs et al. (8)
and was also reviewed in detail by Franklin et al. (9).

Texture analysis. Another general approach to tissue
characterization that has been explored is based on using
the image as a whole and using image processing approaches
to tissue signatures similar to those used for smoothing,
contrast, color enhancement and other cosmetic manipula­
tions of images in general (such as images transmitted from
Mars by the Viking spacecraft). All are aimed at extracting
the most information from the image. After correcting ul­
trasound images for gain control manipulation or attenua­
tion, or both, one asks the computer to evaluate the distri­
bution of gray levels in the pixels within the images (the
so-called "texture analysis") so that information can be
derived about the nature of ultrasound signals returning from
tissues. A recent study by Skorton et al. (10) begins to
explore the capabilities of this approach and technically
reviews texture analysis algorithms, including gray level
histograms, edge counts, gray level run length statistics and
other methods used in these evaluations. These approaches
have the promise of extracting further information about the
nature of the ultrasonic backscatter visualized during echo­
cardiography using advanced signal processing techniques.

Conclusion. Ultrasound technology has advanced sub­
stantially in the past few years. Echocardiography is a safe,
cost-effective and powerful tool for the pediatric cardiologist.

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