Digital Reconstruction and Display of Compound Scan Ultrasound Images

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Abstract—Ultrasonic image reconstruction can be divided into three areas: scan conversion, pixel value assignment (in compound scan imaging), and image display. In scan conversion, the method of allocation of image values in pixels not intersected by scan lines has a dramatic effect on the image quality. Use of an appropriate interpolation scheme can greatly improve image quality and at the same time reduce the required line density and hence increase the frame rate. Peak-detected, averaged, and minimum-detected pixel value assignment lead to dramatically different appearances of the resulting compound scan image. The three methods have different effects on the resolution and on differential diagnosis in clinical conditions. In the display of images formed by the three methods, different postprocessing is required to yield comparable images. This is achieved by an adaptive method based on the required histogram. Results are presented for the reconstruction, assignment, and postprocessing techniques of experimental target studies and clinical data obtained with the U.I. Octoson. The effect of these processes on the appearance of clinical images and application of the techniques to produce image enhancement and differential diagnosis are demonstrated using clinical images of the abdomen.

I. Introduction

Digital techniques have had a profound impact on the capabilities of pulse-echo ultrasonic imaging. The advantages normally claimed include improved stability of imaging properties and control of the signal processing techniques used. However, the use of digital techniques has also changed some of the basic limitations inherent in ultrasonic pulse-echo imaging implemented in analog form.

Scan conversion is the process of transforming data contained in the echos along an ultrasonic line of sight into a set of intensity values at their appropriate positions in an image plane. In an analog system the visual quality of the image depends on having a sufficient number of ultrasonic lines that the gaps between them are unobtrusive in the final image. Thus, the ultrasonic line spacing is approximately equal to the display line width. Using digital techniques interpolation can be used to fill the gaps between data lines. In most systems this is implemented during display by horizontal interpolation between valid data values. By applying the sampling theorem to the spatial frequency distribution of the ultrasonic beam, better approximations to the intermediate values may be obtained by appropriate two-dimensional interpolation. This interpolated scan conversion allows data line spacing to be increased considerably leading to a reduced scan time for static scans and increased frame-rate for real time systems, while retaining all the resolution available from the ultrasound system.

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Pixel value assignment becomes important when compound scan images are reconstructed. Compound scan images contain data from a number of overlapping simple scans. In this case ultrasonic lines of sight may intercept at an image data point. The final value at the data point is a combination of data obtained from the different scan directions. In analog systems the combining process was defined by the available device. Open shutter film recording during the programs of a scan provided an additive process, bistable storage tubes provide thresholded peak detection. Analog scan converters while essentially peak detecting also provide some additive effect at lower intensities. Using digital processing any of these methods may be used and other methods, such as minimum detection, are also possible.

Echoes from tissue consist of strong signals from lines of sight at right angles to surfaces, medium echoes from discrete scattering reflectors from all angles or specular reflectors at nonnormal incidence, and smaller echoes from the distribution of scatterers within the tissue parenchyma. Using compound scanning and the various methods of pixel value assignment, each of these reflector types can be differentiated, thus increasing the diagnostic information.

Image display using analog techniques allows limited smoothing of the data, poor grey scale control, and limited interpolation. Using a digital system, well-defined smoothing operators may be used, operating either on a line-by-line basis or using data within a region. During the display process interpolation can be carried out along the display lines and additional display lines may be obtained by interpolating between data lines. Control of grey scale range may be achieved by a preset compression characteristic or by an adaptive system, based on the desired amplitude histogram in the final image.

The availability of suitable digitally recorded data has encouraged the development of a number of digital analysis techniques which have been lumped under the general heading of "Tissue Characterization" [1]. Perhaps a more descriptive title for those systems used in association with pulse-echo imaging is "Quantified Echography." The techniques seek to obtain information concerning the type or pathological state of tissue. They use a variety of techniques including spectral analysis, pattern recognition techniques, and ray path analysis to quantify such parameters as attenuation and its frequency variation, scattering characteristics, and speed of propagation of ultrasound in tissue. In this paper, improvements in ultrasonic visualization due to digital techniques are described and illustrated using data obtained using the U.I. Octoson.

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II. Equipment

The data used in this paper were obtained using the U.I. Octoson [2] interfaced to a Perkin Elmer (formerly Interdata) Model 85 computer, with a Biomation 8100 transient recorder for data digitization. The received echoes with 3 MHz center frequency are subjected to time gain compensation, logarithmic amplification, full-wave detection, and low pass filtering at 500 kHz. They are then digitized at a sampling rate of 2 MHz and stored together with the relevant deflection information on a 67 Mbyte removable disk drive. Scan conversion and simple or compound scan reconstruction are carried out as a separate procedure at the completion of the data acquisition phase. Display is carried out using a custom-built video memory with 512 X 512 pixels and 8 bits of data of which only 6 bits are used for video data output.

III. Scan Conversion

The initial approach in digital scan cenversion was to plot the locus of the ultrasonic line across the image plane and assign a value from the ultrasonic echo data to each pixel intersected by the ultrasonic line of sight [3]. Distinction was made between pixels intersected at horizontal or vertical boundaries and a number of strategies were used in selecting which echo data sample to allocate to the pixel. The procedure is repeated for the next data line which is displaced laterally by a small distance from the previous one. This procedure relies on the displacement between data lines being sufficiently small that there are no pixels which are "missed" between the lines of sight. If there are a few such pixels, then they give rise to an interference pattern called a Moire pattern, which is quite annoying to the eye and obscures the image [4]. As the line spacing is increased so that the distance between lines of sight exceeds the pixel size, individual data lines are seen with dark spaces between them and the image is further obscured. This problem is normally solved by filtering operations carried out along horizontal and/or vertical lines in the image plane, often "on the fly" during the display operation.

An alternative approach [5], [6] is to treat the ultrasonic lines of sight as samples of the value of a process in a plane and to resample the data on a rectangular grid to form an image. Thus, instead of tracking along the ultrasonic scan lines and locating the pixels intersected, one operates on all the pixels which appear in the region between two adjacent lines in the same scan and on the basis of the data contained in the two lines assign a value to each pixel in the region. These values may be assigned using either the nearest neighbor or the data interpolation approach. The problem is essentially equivalent to that encountered in back projection or Fourier plane reconstructions of computed tomography (CT) images [7], [8].

The sampling theorem states that provided a process is sampled at a frequency twice the highest frequency present in the process, the process can be regained precisely from the samples. The method required to compute the exact values of the process between the sample points is the sinc function or $(\sin x)/x$ interpolation. However, this type of interpolation requires samples away from the area of interest to do the interpolation, and requires a large amount of computing. Provided

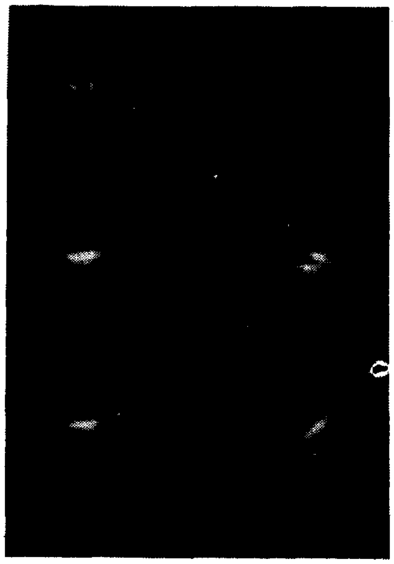


Fig. 1. (a) Two point targets scanned from different directions, using every fifth line from the U.I. Octoson. (b) Linear interpolation along rows of the image, as is commonly used in commercial units. For the left hand target with the vertical beam the result is satisfactory but for the oblique beam interpolation fails. (c) Two-dimensional interpolation gives a good representation of the blur function.

the process is sampled at a greater rate than given by the sampling theorem, classical polynomial interpolation algorithms or "splines" may be used. These include a cubic function using four samples, quadratic using three samples or linear using only two adjacent samples. A frequency domain description of the operation of these classical polynomial interpolation techniques as compared with $(\sin x)/x$ interpolation is available in the literature [9].

The U.I. Octoson performs a compound scan by up to 8 transducers with an angular step between data lines of 0.1°. At the focal distance of 30 cm the distance between scan lines is 0.5 mm. The envelope detected ultrasonic echo signals are digitized and the echo data samples and associated deflection informations are recorded on disk. The sampling rate of 2 MHz fixes the time increment between samples along the one data line at 0.5 μs and the distance between samples along the line at 0.4 mm. The length of the ultrasonic pulse is 1.5 μs corresponding to 1.2 mm and the 6 dB beamwidth at focus is 3 mm and the 20 dB beamwidth is 8 mm. Thus there are at least two time samples in the impulse response function along the beam and 8 samples across the beam width at the focus and more elsewhere. This suggests that fewer data lines may be used provided appropriate two-dimensional interpolation processing is used.

Real-time scanners tend to have reduced line numbers to allow an increased frame rate. For an image consisting of 128 lines and 20 cm depth range the maximum frame rate fixed by ultrasonic speed considerations is 30 frames per second. The attainable frame rate is somewhat less than this to allow for attenuation of the previous pulse, and switch over times for the new transmitted pulse. To attain a higher frame rate, either the number of lines or maximum depth must be reduced. For a linear array scanner, typically 100 mm long, the line spacing for 128 lines is 0.78 mm. For a 90° sector scanner, the line spacing at half depth is 1.2 mm. For a typical im-

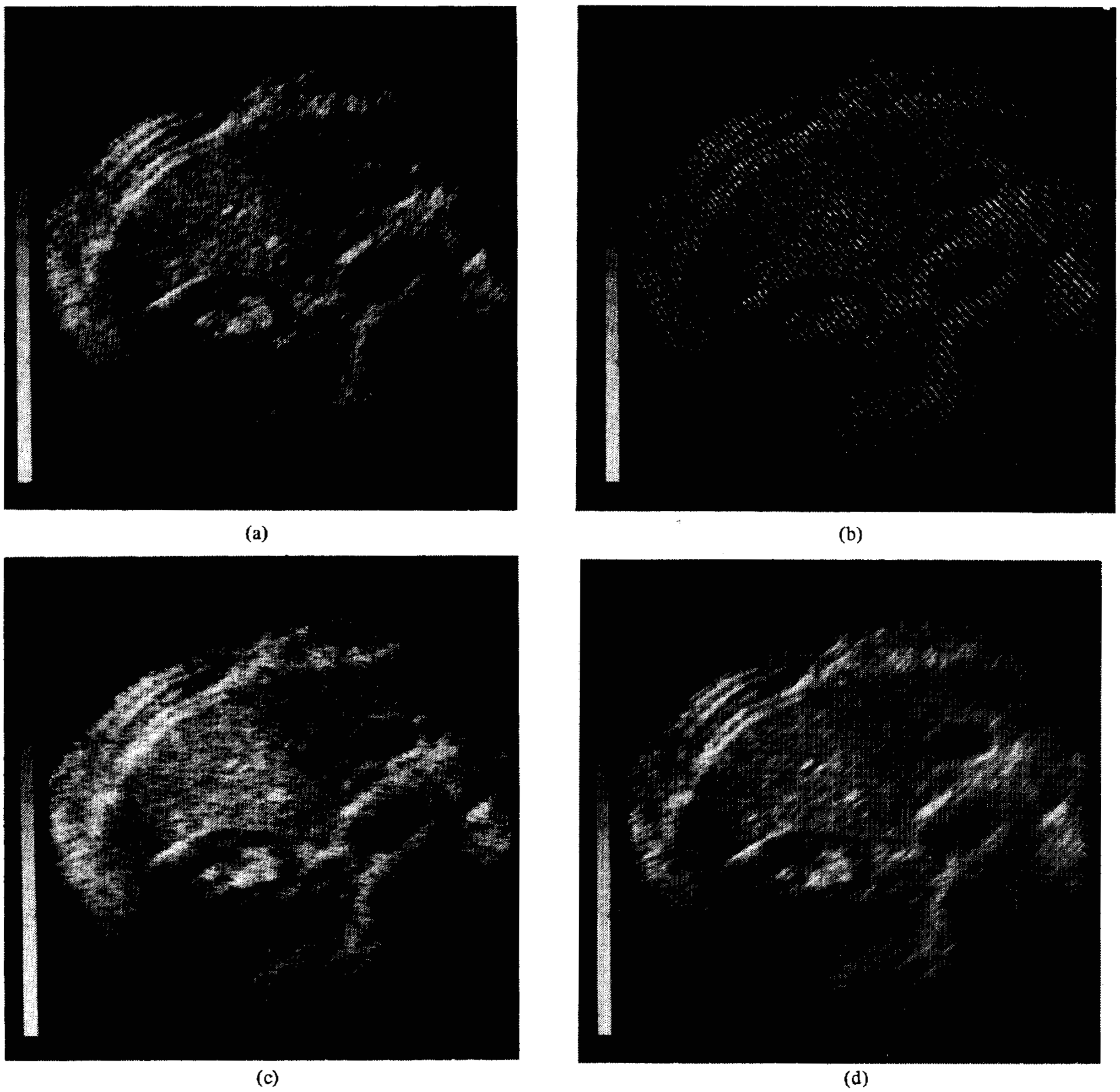


Fig. 2. Sector scan of a normal abdomen. (a) Normal scan with all lines of sight used. (b) Every 4th line used. (c) Linear interpolation of (b). (d) Two-dimensional interpolation. The image of (d) loses little in detail or appearance compared with (a).

age of 512 X 512 pixels representing a 20 cm square in tissue, the pixel size is 0.4 mm, and missed pixels are prevalent.

As the extent of the blur function is different along and across the beam, the highest spatial frequency is also different and thus the required spatial sampling frequency changes. Along the beam the samples must be close together, while across the beam a wider sample spacing is sufficient to convey all the information. However, this information can only be regained by appropriate interpolation. If advantage is taken of the differing sample spacing requirements during acquisition the interpolation must be done at a fixed orientation with respect to the long axis of the blur function, or the axis of the beam. In linear array real-time scanners the beam axis is always vertical and perpendicular to the TV raster lines. Conventional interpolation techniques operate along the TV raster lines, which coincide with the long axis of the blur function formed by the ultrasonic beam width, and thus are appropriate for image interpolation. The U.I. Octoson, static, and real-time *B*-Mode sector images contain oblique ultrasonic lines which are not perpendicular to the TV raster lines, and in these cases horizontal interpolation between scan lines has the effect of degrading the portion of the image formed by oblique lines as the line spacing increases.

The details of the linear two-dimensional interpolation scan conversion algorithm used are published elsewhere [6]. Its effect compared with the conventional methods is shown by computer reconstructions of recorded data in Fig. 1. Fig. 1(a) shows an enlarged view of a conventional reconstruction of a sector scan of a line target perpendicular to the scan plane, with only every fifth Octoson line of sight used. The lines of sight are about 4 pixels apart, and the target is scanned with beams which are vertical and at an angle of 45° to the vertical image direction. Linear interpolation along the horizontal line as shown in Fig. 1(b) increases the size of the blur function in the horizontal direction on the image, which has a component in the axial direction of the beam. This method works correctly for the vertical beam axis case but fails with the inclined beam. The axial resolution is impaired and the edges of the image become jagged. The interpolation scan conversion shown in Fig. 1(c) provides a smooth image with no degradation in the axial direction, for both the vertical and inclined beams.

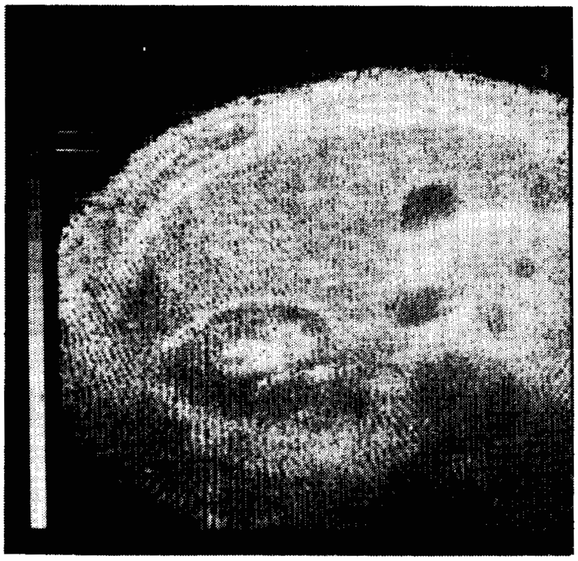
Fig. 2(a) shows a sector scan of the abdomen using all scan lines acquired by the Octoson. The pixel size and scan line spacing are chosen so that there are only very few pixels not intersected by a line of sight. Fig. 2(b) shows the same scan as Fig. 2(a), but using data from only every fourth line of sight. The image is reconstructed at 256 × 256 pixels, on a field size of 144 mm with a pixel size of 0.56 mm. The spacing between lines of sight is approximately 2 mm or about 4 pixels. The linear interpolation along rows commonly used provides a horizontal smearing effect (Fig. 2(c)). The interpolation scan conversion (Fig. 2(d)), produces an image which is only marginally inferior to the original sector image of Fig. 2(a). Thus an equivalent display can be produced with only one quarter of the data needed using conventional scan conversion.

The procedure may be applied to compound scan data. In this case the interpolation is carried out between adjacent ultrasonic lines of sight from within each simple scan. The values obtained from overlapping simple scans are combined as described later to obtain the final image. This result is illustrated in Fig. 3.

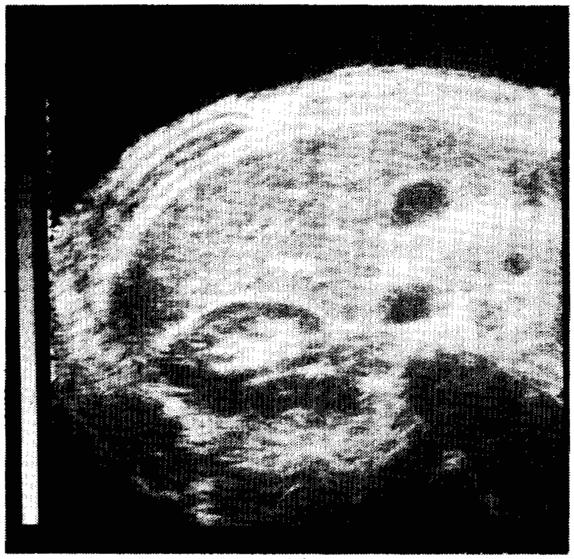
Conventional scan conversion moves the maximum signal value obtained within a pixel to the middle of the pixel. If the maximum occurs near a pixel edge, then it may be detected into both pixels and cause a broadening of the structure, thus reducing the resolution. Interpolation scan conversion also provides a more spatially accurate image, which becomes important when shifts in sector data viewed from different directions are used to compute apparent sound speed in overlying tissue [10].

In a clinical system, the number of ultrasonic lines of sight used is set by the resolution of the display and the image quality required. As a higher frame rate is desirable, the number of lines is reduced, and interpolation used to provide the additional data points at a cost of some image degradation for oblique lines. The interpolation scan conversion system described here allows the use of fewer scanned data lines without any image degradation because it carries out its interpolation in the directions in which the validity conditions for the sampling theorem are fulfilled. The use of the interpolation scan conversion technique makes it necessary to revise the concept of the relationship between ultrasonic line density and image quality. It has been shown that providing the sampling theorem is satisfied and enough lines of sight are obtained so that linear interpolation provides accurate representation of the beam, additional lines of sight do not change the appearance of the image. Hence the traditional trade-off between image quality and frame rate and its implied limit on the usable maximum frame rate must be reassessed in terms of ultrasonic beamwidth rather than number of scan lines.

To achieve the data rate improvement in real-time scanners, it is necessary to implement the necessary interpolation in real-time. A "simplified" two-dimensional hardware interpolation system has been implemented on a 64 line phased array sector scanner [5] but no details are available. Direct implementation of the two-dimensional interpolation algorithm has technological problems. Preliminary studies indicate that such an approach can be implemented with current technology using an array of interpolators operating on different parts of the data in parallel, but the approach appears to be uneconomic.



(a)



(b)

Fig. 3. Compound scan of same section as Fig. 2 with every fourth data line and (a) linear interpolation and (b) two-dimensional interpolation, regaining all the detail of the original scan using all lines of data.

IV. Pixel Value Assignment

In the formation of a compound scan ultrasonic B-mode pulse echo image, the individual sector scans from different transducer positions overlap. Where two ultrasonic lines of sight intersect within a pixel the value to be assigned is determined from the two lines of sight. There are a number of alternative methods by which this assignment may be made. In the past the method was defined by the technology used in the image storage device. Initially direct-view storage tubes were used and these operated in the bistable mode. Any signal above a threshold level was written at full brightness. The grey-scale display technique was originally implemented using a time exposed film to provide adequate storage of the grey-scale echoes [11]. The echo level displayed was the sum of all the echo levels received at each point. The early analog scan converters followed the film storage technique by operating in the integrate mode. In this mode image quality is critically dependent on the evenness of the scanning motion used by the operator and the amount of compounding in compound scans. In an effort to remove this influence scan converters were later used in the equilibrium writing or peak detection mode in which the brightness of each point was proportional to the size of the maximum echo received at that point. Using a digital system the method of assignment of pixel data from values obtained from different directions may be selected arbitrarily to achieve different imaging properties in the final reconstructed image.

Three methods of pixel value assignment have been evaluated, peak detection, minimum detection, and averaging [12], [13]. The peak detection mode corresponds to the standard scan converter image forming system. The largest echo found within each pixel is used as the final value. In minimum detection the maximum echo along the data line within a pixel boundary is detected. This is compared with the current value for that pixel and the minimum of these two values is stored in the memory. The averaged mode approximates that used in the open shutter film store system of image formation. In this mode each echo sample within the particular pixel boundary is added to the existing value associated with the pixel. At the conclusion of the reconstruction the data is normalized by dividing the total in each pixel by the number of accesses to provide the average echo level.

Fig. 4 illustrates the three pixel value assignment modes on a point target using data acquired by scanning a line at right angles to the scan plane. Peak detection yields the familar star shaped pattern. The star shape is due to the beamwidth effect as the echoes are written from different angles. Minimum detection can be thought of as the inverse of peak detection in which the image starts off as completely white and pixels are darkened as smaller echoes are obtained from these regions. The pixels which remain bright are those in which the echoes in the various sectors completely overlap. The beamwidth effects are removed since where the beamwidth is written for one transducer, there is no echo from others. Thus, the apparent lateral resolution on point targets is approximately equal to the axial resolution. However, this result does not carry over by superposition from point targets to echoes from tissue as the imaging system is not linear. In the averaged mode, the echo from the center of the star shaped pattern is accumulated and becomes much larger than that of any of the nonoverlapping beamwidth echoes and there is then an improvement in the apparent resolution for point targets over the peak detected case. Since averaged reconstruction is a linear process, this would be expected to carry over to the case of distributed targets.

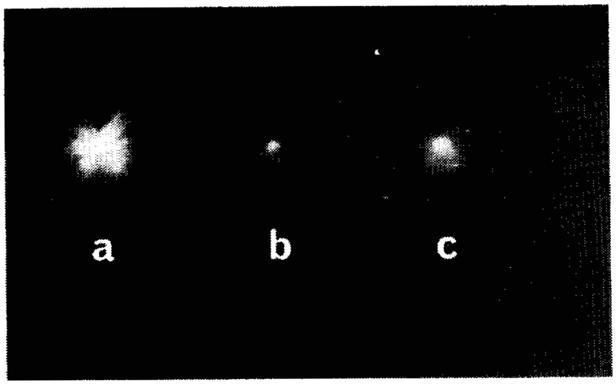


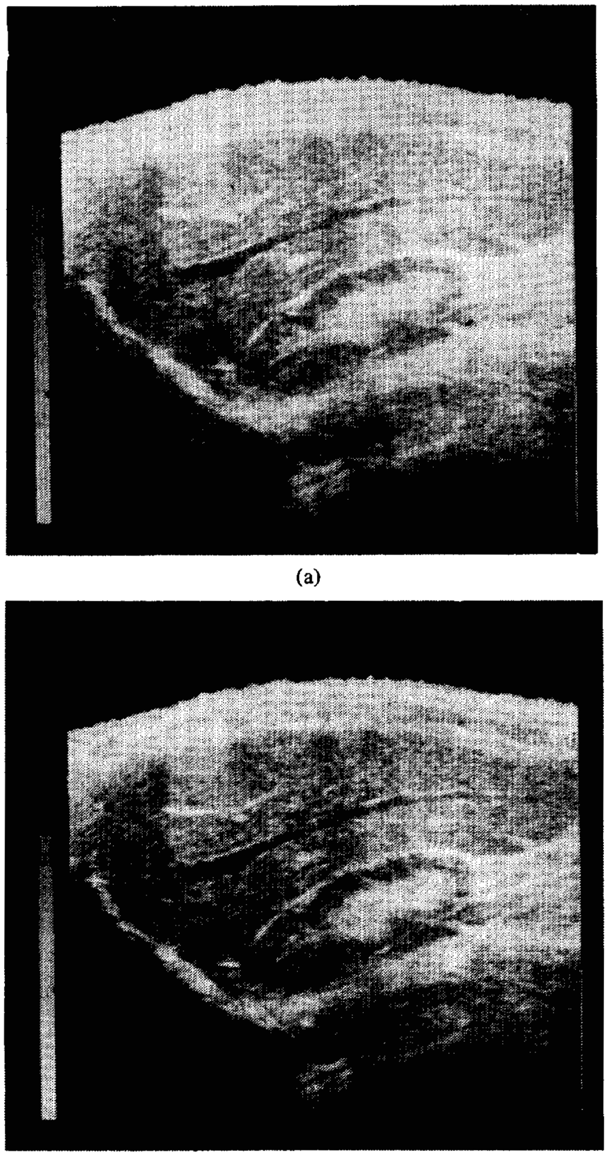
Fig. 4. Point target compound scan data reconstructed. (a) Peak detection. (b) Minimum detection. (c) Averaging.

V. Image Processing

With the flexibility afforded by digital systems, the display device can compensate some of the factors in the data which in analog displays lead to a degradation in image quality.

It is customary in ultrasonic pulse-echo systems to apply nonlinear amplification to the received echoes. The dynamic range of echoes greatly exceeds the available range of the display and decisions have to be made about the mapping characteristic from echo size to intensity level on the screen, to form the so-called grey-scale display [11]. The nonlinear function may be logarithmic, but more often has a more linear characteristic at lower echo levels and a more limiting characteristic for the high-level specular echoes. With a characteristic of this type, care must be taken with the setting of the overall system sensitivity, to ensure that the transition between the linear and limiting portions is at an appropriate level. This level varies from patient to patient and even within the one examination as the efficiency of coupling to the patient and the attenuation of overlying tissue changes for different sections. When a large number of images are reconstructed from prerecorded data it is attractive to have some means to set the grey-scale balance of each image automatically.

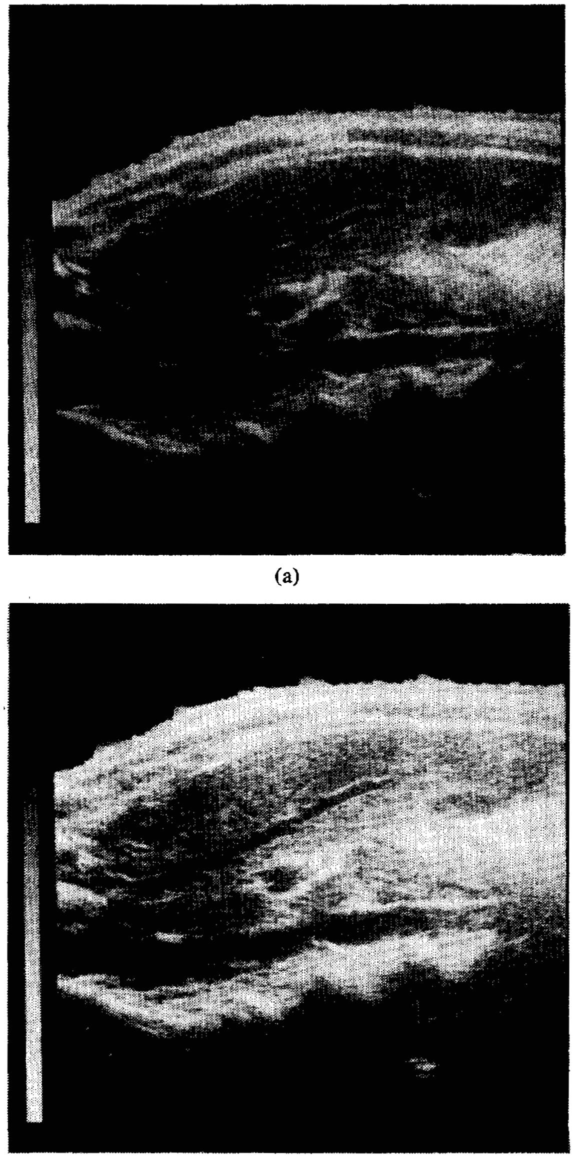
When the different pixel value assignment techniques are applied to the same data the distribution of image intensity varies widely. For effective comparison of the properties of the various techniques it is desirable to have images whose grey-scale distribution is comparable, so that the intrinsic imaging properties become apparent. One method of achieving this is to subject the data to an adaptive nonlinear process to adjust the intensity histogram to a desired characteristic. The characteristic normally used for this purpose is a constant value; that is, there are equal numbers of pixels containing each intensity level, and the process is termed histogram equalization [14]. This procedure has been found optimal for many classes of image including aerial photographs, landscapes, and portraits. However, when applied to ultrasonic pulse-echo images the results has too much contrast and too great a proportion of highlight echo area and some histogram shape other than flat is required. A survey of a panel of experienced ultrasonic image interpreters was used to determine the appropriate histogram for different clinical situations. For peak detected and averaged images of the pregnant uterus and abdomen it was found that the desired histogram is constant up to half inten-



(a)

(b)

Fig. 5. Example of image display with the image processed for (a) flat; and (b) constant-sine histogram.



(a)

(b)

Fig. 6. Reconstruction of the same data. (a) 256 X 256 pixels with linear interpolation between rows and columns to obtain a 512 X 512 pixel display. (b) 512 X 512 pixel reconstruction.

sity and then decreases with a sine curve to zero at full intensity [13]. This histogram curve is referred to as constant-sine. For breast examinations and abdominal sections with a large area of high level echoes this processing still produces an excessively bright image, and a cosine curve over the whole intensity range was found most satisfactory. For minimum detection, fewer bright areas are required and cosine is used for abdominal and obstetric images and a cosine squared curve has been found most desirable for breast images. An example of flat and constant-sine histogram shaping on an abdominal image is shown in Fig. 5.

The required pixel size is set by the desired clarity of the ultrasonic image, and the resolution inherent in the ultrasonic data. For typical ultrasonic images the underlying resolution is such that 256 X 256 pixels are sufficient. However an image with this pixel density has a "blocky" appearance, and the perceived quality of the image can be improved by interpolating the data up to 512 X 512 pixels by linear interpolation between points along each row and interpolating point by point between rows. Fig. 6 shows the comparison of the same data reconstructed at 256 X 256 and linearly interpolated, and reconstructed directly at 512 X 512 pixels. There is little discernible difference between the two images.

Conventional digital display techniques require 5 or 6 bits of resolution of displayed data to avoid problems of "contouring" in regions of the image in which the intensity varies slowly with displacement. A method of avoiding this problem in data with low dynamic range (number of bits) is to add a small amount of random notes in the display process. This breaks up the "contours" and improves the appearance of the image. In the display of ultrasonic images, the data already includes a noise signal due to the phenomenon of speckle in the imaging characteristics of the system. Thus for display of ultrasonic image, the degradation when going from 6 bits to 4 bits

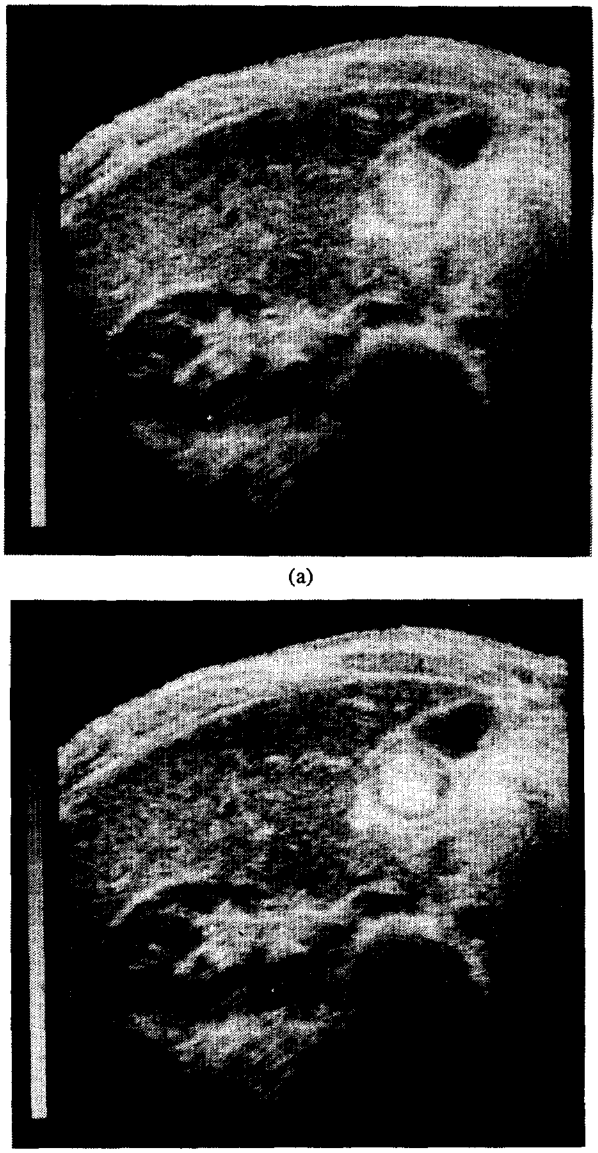


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Fig. 7. (a) Image displayed with 6 bits. (b) Image displayed with 4 bits of intensity resolution.

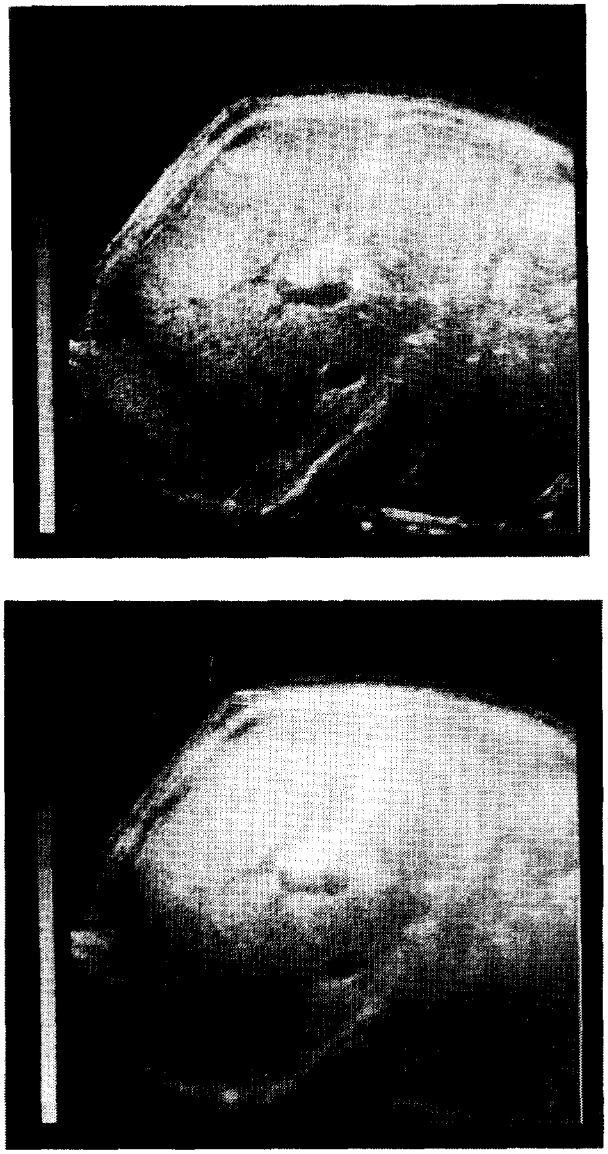


Fig. 8. (a) Example of peak detected. (b) Example of averaged reconstruction of data from a normal liver. Latter shows improved spatial and amplitude resolution and reduced speckle.

of resolution is scarcely visible (Fig. 7). It should be emphasized that this result is only true for the final display. Any processing, particularly amplitude windowing of the reconstructed image, requires a greater resolution so that the final image to be displayed after postprocessing still has a resolution of 4 bits.

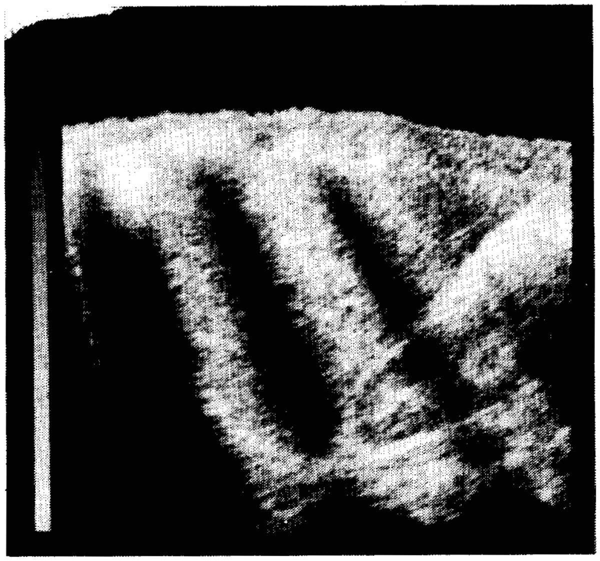
All the figures in this paper, unless specified otherwise, use interpolated 256 × 256 pixel data with a dynamic range of 6 bits and constant-sine histogram shaping.

VI. Results on Clinical Data

Peak detection is the method normally used in pulse-echo visualization. Although it gained favor due to reasons associated with the limitations of available technology, it has a number of features which make it useful as a matter of choice. Since a strong echo obtained from any direction will be retained at full strength in the final image, it tends to emphasize specular reflections seen from only a single direction such as outlines of organs and structures. Detail on layered structures can best be displayed by examining the structures at normal incidence with a reduced gain setting. A peak detected compound scan gives a comparatively complete image of the abdominal organs (Fig. 8). Peak detection retains the beamwidth echoes at their full size. An area with low reflectivity such as the lumen of a vessel or duct may well be obscured by the large amplitude echo from the side of the beam reflecting from an adjacent strong reflector. Thus fine detail in lower amplitude echoes is easily obscured.

Compounding and peak detection reduce the obscuring effect of rib shadowing in examination of the upper abdomen, since beams from other directions effectively fill in the shadow behind ribs (Fig. 9).

Minimum detected reconstruction is essentially a subtractive technique which stores the smallest echo received from each pixel. Thus the large specular echoes received from only one



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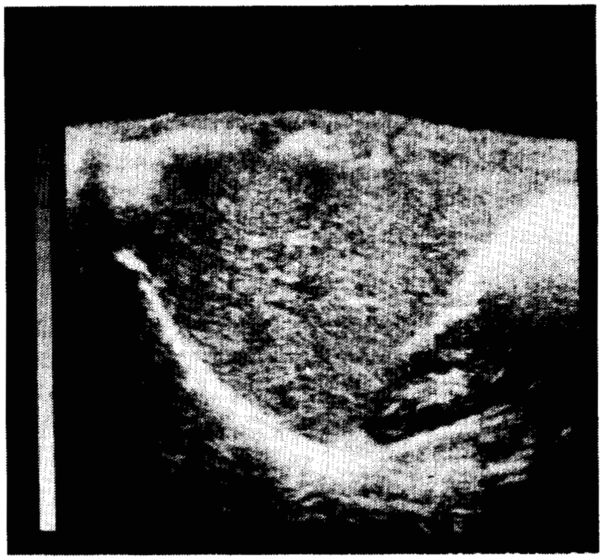

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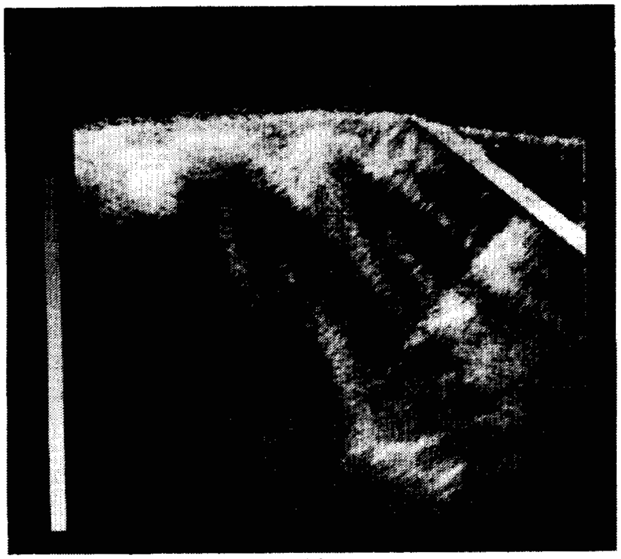
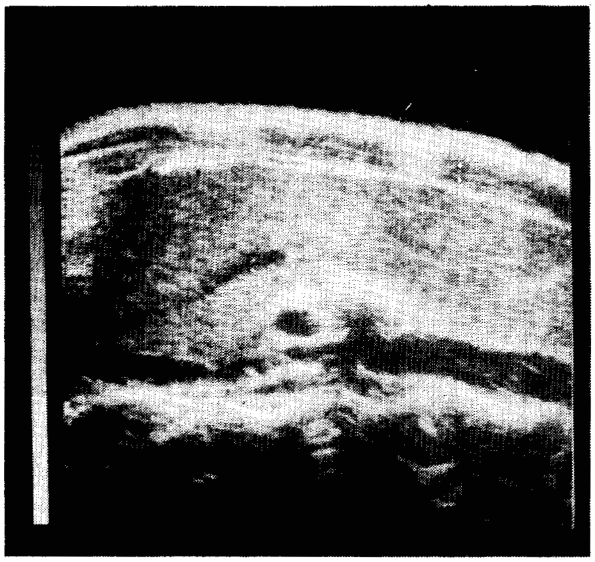


(c)

Fig. 9. (a) Sector scan. (b) Compound peak detected. (c) Compound minimum detected reconstruction of a liver section showing the way shadows from different directions combine to block out the image.



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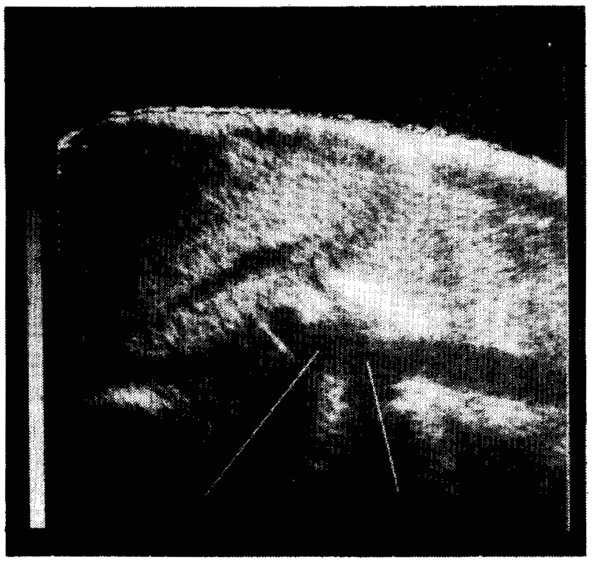

(b)

Fig. 10. Example of use of minimum detection to locate a shadowing region in tissue. (a) Peak detected compound scan of a longitudinal section of the liver. (b) Minimum detected two-transducer scan showing the shadowing effect of a stone in the cystic duct. Intersection of the two shadows fixes the position of the stone.

direction are deleted from the image, which is thus composed predominantly of scattered reflections. Structures which return strong reflections in all directions are retained as strong echoes in a minimum detected image.

If a region is not displayed in one of the sectors making up the compound scan that region is removed from the final echogram. Rib shadows (Fig. 9(b)) or shadowing from the fetal spine from different directions superimpose to effectively blank out the entire image. Shadowing tends to limit the effectiveness of minimum detected compound scans. The phenomenon can be put to use to act as a pointer to a shadowing region. When a shadow is cast by a discrete region such as a stone in the cystic duct (Fig. 10) the individual shadows from the two sector scans used in minimum detection combine to form an arrow head with the shadowing structure at its point. When a shadow is observed in a single sector scan the exact

(a)

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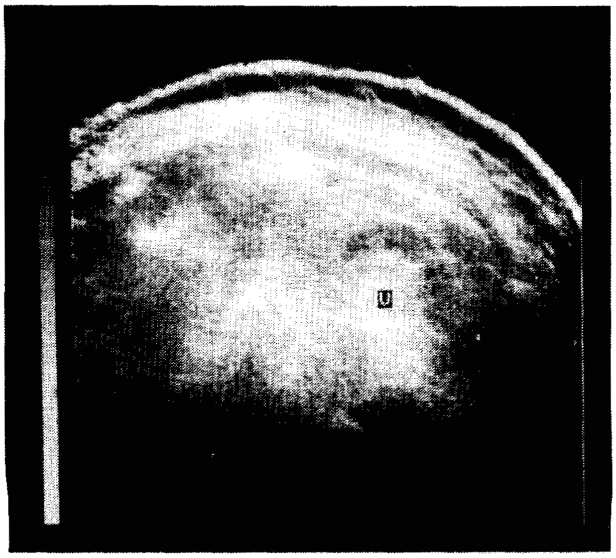
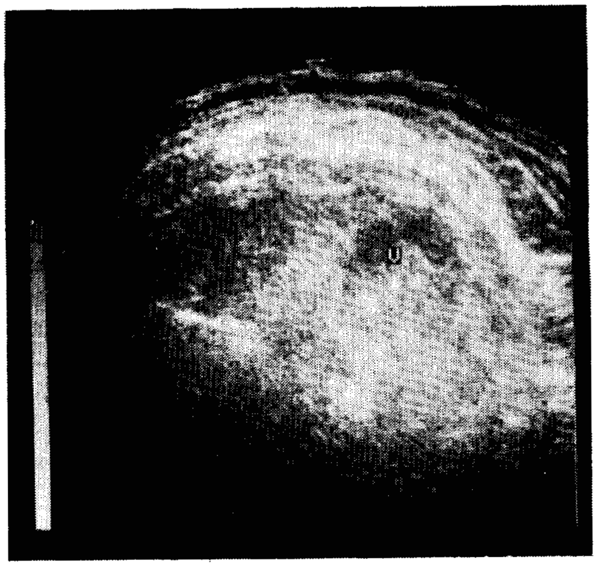


(c)

Fig. 11. Images from kidney containing a stone. (a) Not seen in the peak detected image. (b) Displayed in the minimum detected image. (c) Displayed in the averaged image.



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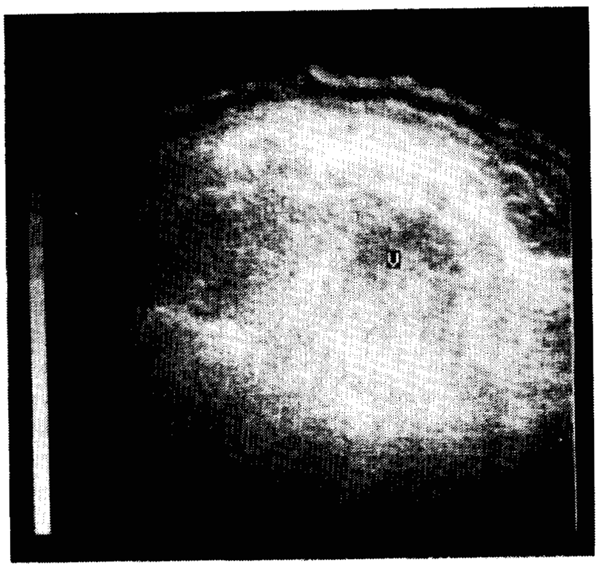

(b)

Fig. 12. A counter example showing echoes resembling stones on (a) peak detected images which are not seen on (b) averaged image.

depth of the shadowing structure is not always apparent. This is particularly so when there are some echo-free areas in the region of the shadow. A small number of sectors, say two or three, reconstructed in minimum detected mode will readily point out the true position of the source of the shadow by the intersection of the shadows produced. This technique will also demonstrate whether the shadows seen on individual sectors originate at a single region, or from a number of them.

A major feature of averaged reconstruction is the improvement in resolution obtained. This result was predicted very early in the development of ultrasound, but it had not been attained previously due to limitations in technology. The degradation evident in peak detected compound scan images due to beam-width broadening is greatly reduced. In an echogram of the liver (Fig. 8(b)) the averaged reconstruction shows greatly improved detail in the intrahepatic structures. The textural appearance of the abdominal organs is markedly smoothed. Thus the resolution improvement observed under averaged reconstruction for point targets is also found in tissue. This is to be expected as averaged reconstruction constitutes a linear image forming system.

Reflection from specular reflectors are displayed less prominently in averaged reconstruction. Strong reflections from a single direction, which are not obtained from other directions, are preserved at full strength in peak detection, but are proportionally reduced compared with echoes from scatterers when averaged. However, echoes from strong reflecting scatters are enhanced by averaged reconstruction. In a study involving the location of stones within the calyceal system of the kidney, the problem rests in identification of small stones within the strongly specularly reflecting surfaces of the calyceal system. It has been seen (Fig. 11) that averaged and minimum detected images locate strongly scattering discrete reflectors not apparent on the peak detected image. On the other hand, echoes having a similar appearance on peak detected images, but not seen on averaged or minimum detected reconstructions of the same data came from a kidney in which no stone was present (Fig. 12).

Each of the reconstruction methods imparts a different appearance to the echogram. Using the conventional technique of peak detection, considerable clinical information is gained by visual assessment of the appearances of detail within the echogram. The availability of different reconstruction methods provides another parameter for visual interpretation of tissues character from the appearance of the echogram.

VII. Conclusion

The application of digital techniques has greatly increased the capability of pulse-echo ultrasonic imaging as a diagnostic technique in medical practice. The usual advantages of digital techniques of improved operator interface, preset control parameters and greater stability of processing and display are realized. However, digital techniques also allow basic changes in image reconstruction and display which were not practical using the analog techniques they displaced. The effect of this is that previous theoretical limitations on equipment performance have been removed by more efficient use of the ultrasonic data. The relationship between number of ultrasonic lines and image quality is modified by the use of more appropriate between-line interpolation techniques. This will have an impact on real-time scanner image quality as the improvements in technology, in either device speed or using greater parallelism with VLSI, allow a real-time two-dimensional interpolator to be built economically. Imaging techniques in compound scanning using different pixel value assignment algorithms allow differential diagnosis to be made between specular and scattering reflectors. Averaging offers improved spatial and echo amplitude resolution and reduction of the speckle effect. Minimum detection allows the location of shadow-producing regions in a single image by triangulation. Image display techniques using the desired histogram as the criterion for a nonlinear amplitude mapping process removes the need for skilled selection of the nonlinear characteristic and sensitivity level on an image-by-image basis.

The ability to record the raw ultrasonic echo data digitally for subsequent use for image reconstruction has the effect of reducing examination time, as at present interpretation of the examination data takes place at least partially during the course of the examination. This has obvious economic effects, but also permits the successful examination of very sick patients who cannot tolerate a prolonged procedure with controlled posture and respiration. Use of a digital ultrasound system as a front-end for a more extensive computer-based system offers the possibility of further developments in *quantified echography.*

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