Range- and azimuth-dependent variability of image texture in two-dimensional echocardiograms

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ABSTRACT Regional two-dimensional (2D) echocardiographic amplitude patterns, or image texture, may be of diagnostic importance. Echocardiographic image texture is due in part to acoustic speckle, a complex pattern of interference of reflections from many small scatterers in tissue. The regional speckle pattern appears to be altered in several disorders associated with abnormalities in myocardial structure but also may be altered by a variety of characteristics of the scanning instrument. We hypothesized that quantitative measures of regional 2D echocardiographic image texture would vary as a function of position in range and azimuth within the field of view, even when imaging a uniform ensemble of scatterers. We tested this hypothesis by imaging a tissue-equivalent phantom with two phased-array scanners and two different methods of digitization. We analyzed the texture in several regions of interest separated in range and azimuth and found significant differences in quantitative texture measures as a function of position of the region of interest in the sector field of view (p values .006 to .0001 by multivariate analysis of variance). We found significant regional variability in texture with both scanners and both methods of digitization. We conclude that regional quantitative image texture in 2D echocardiograms varies as a function of range and azimuth, even when imaging a uniform ensemble of scatterers. This variability is related to several physical and instrument-related phenomena and precludes interpreting all regional texture alterations as indicating tissue structural abnormalities.

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THE CHARACTERISTICS of regional echocardiographic amplitude patterns in B mode ultrasound scans may be of diagnostic importance. In particular, the two-dimensional spatial pattern of gray levels, usually described as image "texture," is used in the analysis of B mode scans of liver, breast, and other organs. Qualitative abnormalities of echocardiographic image texture have also been noted in two-dimensional (2D) echocardiograms, such as the peculiar appearance of echoes returning from the ventricular septum in echocardiograms of patients with hypertrophic cardiomyopathy.

In an attempt to evaluate image texture in B mode ultrasound scans in a more objective fashion, quantitative texture analytic methods have been applied to data from scans of the liver and heart. Analysis of patterns of A mode signals has been successful in identifying abnormal liver tissue. Using video data from 2D echocardiograms, we have identified regions of acute myocardial contusion with a set of classic texture calculations. The use of quantitative echocardiographic texture analysis for tissue characterization is based on the assumption that alterations in image texture are due to alterations in tissue structure. However, image texture in B mode scans is mainly produced by constructive and destructive interference of echoes arising from many small scatterers — a pattern referred to as acoustic speckle.

Further, B mode ultrasound images are produced by a series of filtering and detection processes in the ultrasound scanning instrument. Thus, the precise relationship between tissue structure (i.e., the scatterer ensemble) and image texture is by no means clear or direct. In fact, on the basis of computer modeling, Flax et al. have shown that B mode image texture is the complex result of a variety of physical and instru-
ment factors. These investigators have predicted that texture would vary in different regions of a B mode scan, even when imaging a statistically uniform ensemble of scatterers. On the basis of this previous work and on considerations of image formation in real-time echocardiograms, we hypothesized that regional image texture would vary within the sector field of view of 2D echocardiographic images. Specifically, we sought to determine whether quantitative texture calculations would vary as a function of range (distance from the transducer) and azimuth (lateral offset from the centerline of the scan) in 2D echocardiographic images of a uniformly scattering, tissue-equivalent phantom.

**Methods**

**Experimental target (phantom).** The scattering target we used was a commercially available tissue-equivalent phantom (Model 412; RMI Inc., Middleton, WI). This device consists of a plastic block containing agar gel in which are embedded graphite particles with an average size of 7 μm. This graphite-in-gel material has ultrasound velocity and attenuation characteristics similar to those of soft tissue. The graphite particles are distributed uniformly and produce a texture pattern similar to that of soft tissue on B mode scans. The phantom was imaged through a Mylar window by use of a standard acoustic gel as a coupling medium.

**Echocardiographic examination and recording.** We performed 2D echocardiography using two phased array instruments: scanner I (Toshiba SSH-10A Sonolayergraph) operating at 2.4 MHz and scanner 2 (Diasomics V3400R) operating at 2.25 MHz. In both cases the scanner overall gain and time-gain compensation controls were adjusted to produce an image of the phantom material with apparently uniform regional brightness (gray level). The echocardiographic transducer was held rigidly in place with a clamp to avoid motion artifacts in the images. With scanner 1 we obtained five images using each of two scanning windows in the phantom (designated "top" and "side" [diagonal] windows, figure 1) to determine any angle dependency of the displayed image texture. With scanner 2 we obtained five images from only the top imaging window. Images were recorded and digitized differently with the two scanners.

**Image digitization.** Images from scanner 1 were photographed from the primary cathode ray tube with a digital freeze-frame system. Photographs were obtained with Polaroid 665 positive/negative film (exposure of 5 sec., f-stop of 4.5) and negatives were cleared with sodium sulfite. These negatives were placed on a light table and digitized with a video camera (Hamamatsu C-1000), resulting in a matrix of 256 × 256 pixels with eight-bit (256) gray level quantization. The images obtained in this manner were digitized at identical magnification and under identical conditions of ambient illumination.

Images from scanner 2 were acquired directly in digital form with a data transfer system developed with the manufacturer. In this system the image was stored, after analog-to-digital conversion, in a buffer memory in the echocardiograph. The data were stored in a "pseudopolar" format, with 128 lines of data, each containing 470 sampling points (pixels). The digital data were transferred from the buffer memory to a PDP 11/34 computer with an RS232 interface. These digital images were "reconstructed" into a 256 × 256 pixel rectangular format for display and analysis. The echocardiographic amplitude data were quantized to eight bits, or 256 levels.

**Quantitative texture analysis.** The digitized images were displayed on a 256 × 256 pixel image display system (Ramtek 9050) interfaced to a PDP 11/34 computer for quantitative image analysis. In each image several regions of interest were identified, varying as a function of range (distance from the transducer) and azimuth (offset from the centerline of the scan). In the images from scanner 1 obtained from the top acoustic window, eight such regions of interest were chosen (figure 1, A), and in the images from scanner 2, three were chosen (figure 1, B). In the case of scanner 2, the regions of interest were separated only in range.

The data within each region of interest were analyzed with several quantitative texture measures that can be categorized into three groups: (1) edge count, 7 (2) gray level run-length statistics, 12 and (3) gray level difference statistics. 13 These three groups of texture calculations are briefly described below; specific algorithms are reported elsewhere. 7, 12-14

The edge count consists of identifying the heterogeneity of the gray level data within each region of interest by a count of edge points in the region of interest. This was accomplished in two stages. First, a two-dimensional gradient edge operator
(Sobel operator) was used to identify potential edge points in each region of interest by measuring the change in gray level in vertical and horizontal directions across each pixel. The Sobel operator uses a $3 \times 3$ pixel window or mask, which is systematically moved across all portions of the image. If the combined vertical and horizontal gray level gradient exceeds an operator-identified threshold, then the pixel of interest (center pixel of the $3 \times 3$ mask) is designated as an edge pixel. A global gray level threshold, determined from the global gray level histogram, was used to identify edge pixels throughout the image. This process resulted in an "edge image." The number of edge pixels in each region of interest was then counted and tabulated.

Gray level run-length statistics are based on the concept of a run length as a set of consecutive pixels having the same gray level value or having gray level values within a specified range. Five run-length parameters were calculated: (1) long run emphasis (higher values indicate longer average run lengths), (2) short run emphasis (higher values indicate shorter average run lengths), (3) gray level nonuniformity, a measure of the tonal distribution (smaller values indicate that run lengths are evenly distributed throughout the gray levels), (4) run-length nonuniformity, a measure of the run-length size distribution (smaller values indicate that runs are evenly distributed throughout the run-length groups), and (5) run percentage (ratio of actual number of runs to possible number of runs in the image). Each of these calculations was performed along vertical and horizontal directions within each region of interest.

Gray level difference statistics consist of measuring the absolute difference between the gray level values of two points separated by a fixed distance. We measured gray level differences at separations of 1, 2, 4, and 8 pixels in horizontal and vertical directions within each region of interest. For each region of interest, a tabulation of the differences measured resulted in a "difference histogram" (i.e., the frequency of occurrence of all differences encountered within each region of interest). From the difference histogram, four parameters were derived: (1) contrast, measuring the moment of inertia about the origin of a gray level difference histogram (larger values indicate a shift of the histogram away from the origin), (2) angular second moment, a measure of the uniformity of the difference histogram (smaller values indicate a more uniform histogram), (3) entropy, a measure of the randomness or degree of spread of the histogram (larger values indicate a more uniform histogram), and (4) mean, a weighted average of all difference-histogram entries, measuring the degree of spread away from the origin (larger values imply that more values are concentrated at larger gray level differences).

**Statistical data analysis.** Multivariate analysis of variance
was used to assess the significance of differences between texture measures calculated at the different positions (regions of interest) in range and azimuth and to assess the significance of differences between images. Due to the large number of variables calculated, the required levels of statistical significance were $p = .0025$ (scanner 1) and $p = .008$ (scanner 2), by use of the Bonferroni multiple comparison method.\textsuperscript{15}

**Results**

**Qualitative image texture.** Figures 2 and 3 are representative 2D echocardiographic images of the phantom obtained with scanners 1 and 2, respectively. In both figures a regional variation in textural appearance is exhibited that illustrates the typical pattern we found: the image texture is discrete and “granular” in the near field and becomes progressively smoother and coalescent as we observe regions more distant in range. A similar variation in image texture is noted whether the phantom is scanned from the top (figure 2, A) or from the side (figure 2, B). Therefore, the regional variation displayed in these images is not due to the particular arrangement of the scatterers (graphite particles) within the gel.

**Quantitative image texture.** In table 1 we have listed mean values and SDs of all texture calculations performed for selected regions of the images obtained with both scanners. Significant changes in texture measures were noted for regions separated in range and in azimuth (figures 4, 5, and 6). Range differences were noted in images obtained with both scanners (azimuthal differences were not analyzed for scanner 2). The differences noted in range (figures 4 and 6) were more substantial than those noted in azimuth (figure 5). We found similar range and azimuth variability of quantitative texture measures in images obtained from both top and side scanning windows with scanner 1.

**Discussion**

In this article we have demonstrated that quantitative measures of texture in 2D echocardiographic images vary as a function of range and azimuth even when imaging a uniform ensemble of scatterers. Changes in texture with range were more marked than those with azimuth. The variability with range and azimuth was present in images acquired with both echocardiographs we tested. In this discussion we will review the factors that affect the texture observed in 2D echocardiograms and discuss the implications of our findings with respect to the use of texture analysis to detect tissue abnormalities.

**Factors affecting image texture.** We define echocardiographic image texture as the two-dimensional spatial distribution of echocardiographic amplitudes or gray levels. The texture visible in B mode scans of soft tissue regions, often referred to as acoustic speckle,\textsuperscript{8, 9} is due to constructive and destructive interference of returning ultrasound waves scattered from multiple reflectors that are small compared with the wavelength of the interrogating ultrasound signal.\textsuperscript{10} Qualitative observations\textsuperscript{5, 17, 18} and quantitative analyses\textsuperscript{7} have suggested that texture in 2D echocardiograms depends in part on the microscopic structure of the imaged tissue. These observations form the rationale for the use of 2D echocardiographic texture analysis for tissue characterization. Unfortunately, the characteristics of the ultrasound imaging system also affect image texture. Flax et al.\textsuperscript{10} have demonstrated that B mode ultrasound image texture can change dramatically when different transducers are used. Although one might assume that the fineness of ultrasound image texture would be directly related to the regional resolution–cell size, this is not necessarily the case. Moreover, textural variations in the axial or azimuthal directions are influenced by different phenomena.\textsuperscript{10} The specific signal processing used in many scanning instruments (nonlinear ampli-
### TABLE 1

Texture measures for selected regions (mean ± SD)

<table>
<thead>
<tr>
<th>Measures</th>
<th>Scanner 1</th>
<th>Scanner 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>C₁</td>
<td>C₃</td>
</tr>
<tr>
<td>Edge count</td>
<td>526 ± 52.6ᵃ</td>
<td>423 ± 38.8</td>
</tr>
<tr>
<td>Run Length 0° (horizontal)</td>
<td>0.309 ± 0.015</td>
<td>0.333 ± 0.274</td>
</tr>
<tr>
<td></td>
<td>6.94 ± 0.475ᵇ</td>
<td>9.17 ± 0.814</td>
</tr>
<tr>
<td></td>
<td>150 ± 4.58ᵇ</td>
<td>106 ± 9.88</td>
</tr>
<tr>
<td></td>
<td>126 ± 12.7ᵇ</td>
<td>69 ± 9.5</td>
</tr>
<tr>
<td></td>
<td>0.263 ± 0.015ᵇ</td>
<td>0.200 ± 0.017</td>
</tr>
<tr>
<td>Run-length 90° (vertical)</td>
<td>0.457 ± 0.032</td>
<td>0.472 ± 0.017</td>
</tr>
<tr>
<td></td>
<td>4.85 ± 0.492</td>
<td>4.56 ± 0.286</td>
</tr>
<tr>
<td></td>
<td>206 ± 15.99</td>
<td>205 ± 12.04</td>
</tr>
<tr>
<td></td>
<td>191 ± 29.9</td>
<td>206 ± 20.7</td>
</tr>
<tr>
<td></td>
<td>0.363 ± 0.031</td>
<td>0.384 ± 0.022</td>
</tr>
<tr>
<td>Gray level difference 0° (horizontal)</td>
<td>106 ± 12.4ᵇ</td>
<td>63 ± 7.3</td>
</tr>
<tr>
<td>ΔX = 1</td>
<td>0.0575 ± 0.0044ᵇ</td>
<td>0.0775 ± 0.0045</td>
</tr>
<tr>
<td></td>
<td>2.99 ± 0.0690ᵇ</td>
<td>2.72 ± 0.0601</td>
</tr>
<tr>
<td></td>
<td>0.0336 ± 0.0020ᵇ</td>
<td>0.0260 ± 0.0014</td>
</tr>
<tr>
<td>ΔX = 4</td>
<td>717 ± 112ᵇ</td>
<td>447 ± 59.0</td>
</tr>
<tr>
<td></td>
<td>0.0219 ± 0.0020ᵇ</td>
<td>0.0284 ± 0.0019</td>
</tr>
<tr>
<td></td>
<td>3.96 ± 0.886ᵇ</td>
<td>3.72 ± 0.0641</td>
</tr>
<tr>
<td></td>
<td>0.0851 ± 0.0073ᵇ</td>
<td>0.0665 ± 0.0044</td>
</tr>
<tr>
<td>90° (vertical)</td>
<td>225 ± 36.5</td>
<td>245 ± 15.8</td>
</tr>
<tr>
<td>ΔY = 1</td>
<td>0.0413 ± 0.0030ᵇ</td>
<td>0.0382 ± 0.0010</td>
</tr>
<tr>
<td></td>
<td>3.37 ± 0.0796ᵇ</td>
<td>3.42 ± 0.0330</td>
</tr>
<tr>
<td></td>
<td>0.0468 ± 0.0036ᵇ</td>
<td>0.0498 ± 0.0014</td>
</tr>
<tr>
<td>ΔY = 4</td>
<td>910 ± 161.1ᵇ</td>
<td>1193 ± 173.1</td>
</tr>
<tr>
<td></td>
<td>0.0196 ± 0.0020ᵇ</td>
<td>0.0167 ± 0.0011</td>
</tr>
<tr>
<td></td>
<td>4.07 ± 0.909ᵇ</td>
<td>4.19 ± 0.0627</td>
</tr>
<tr>
<td></td>
<td>0.0959 ± 0.0093ᵇ</td>
<td>0.1132 ± 0.0095</td>
</tr>
</tbody>
</table>

C₁, C₃, R₁, TC, BC = regions of interest (see figure 1); SRE = short run emphasis; LRE = long run emphasis; GLN = gray level nonuniformity; RLN = run-length nonuniformity; RP = run percentage; ΔX, ΔY = separations (in pixels) in horizontal and vertical directions; CON = contrast; ASM = angular second moment; ENT = entropy.

Statistical comparisons: ᵃp < .0025 compared with region C₁; ᵇp < .001 compared with region C₁; ᶜp < .001 for all BC regions compared with TC regions.

tude compression, envelope detection, thresholding) also modifies the displayed image texture. An additional factor that potentially affected image texture in the present study was our use of photographic techniques for image acquisition from scanner 1. Since our results with photographic and direct image acquisition were similar, however, it is likely that the acquisition technique is not a major contributor to image texture. Although a number of authors have proposed theoretical models to explain certain aspects of ultrasound image texture,⁶,⁹ the precise relationship between the characteristics of the imaging system and image texture remains unclear.

Implications of regional variability in ultrasound image texture. Changes in ultrasound image texture have been used to imply the presence of abnormal tissue.¹⁻⁵,⁻⁷,⁻¹⁷,⁻¹⁸ However, as demonstrated in this study, image texture varies with both the axial and azimuthal position of the
scanned object. Thus, it is probably inappropriate to interpret regional alterations of 2D echocardiographic image texture as necessarily indicating abnormalities in tissue structure. What is the contribution of tissue structure to the echocardiographic image texture? How can the contributions to echocardiographic image texture of tissue structure and scanner characteristics be distinguished? This study did not attempt to answer the first question (i.e., the relationship between tissue structure and echocardiographic image texture).

**FIGURE 4.** Texture calculations are shown for two regions separated in range obtained with scanner 1. Significant alterations were found in edge count, in run-length long run emphasis calculated horizontally (LRE - 0 degrees), and in gray level difference contrast calculated for a horizontal separation of 4 pixels (0 degrees, ΔX = 4). C₁ = center region, position 1; C₃ = center region, position 3 (see figure 1, A).

**FIGURE 5.** Texture calculations are shown for two regions separated in azimuth obtained with scanner 1. A significant difference was found in gray level difference contrast calculated for a horizontal separation of 4 pixels (0 degrees, ΔX = 4). No significant difference was found between these two regions in run-length long run emphasis calculated horizontally (LRE - 0 degrees). The difference noted in edge count approached but did not reach statistical significance (p = .0065). C₃ = center region, position 3; R₃ = right region, position 3 (see figure 1, A).

**FIGURE 6.** Texture calculations are shown for two regions separated in range obtained with scanner 2. Significant differences were found in edge count, in run-length long run emphasis calculated horizontally (LRE - 0 degrees), and in gray level difference contrast calculated for a horizontal separation of 4 pixels (0 degrees, ΔX = 4). TC = top center position; BC = bottom center position (see figure 1, B).
eral potential answers to the second question can be envisioned. One approach is simply to compare texture from regions at a similar axial and azimuthal position. We have successfully used this method to detect myocardial contusion with quantitative image texture analysis. Another approach is to "correct" quantitative measures of texture for changes in range or azimuth. For example, run lengths in a particular direction might be normalized by the resolution cell size in the same direction. It might be possible to devise specific quantitative texture measures that are relatively insensitive to changes in range and azimuth and yet are sensitive to pathologic changes in tissue structure. A final approach is to reduce the contribution of the imaging system to the displayed texture. This is referred to as speckle reduction, and several methods, including parallel focusing and compound imaging, have been proposed and are currently under investigation.

In conclusion, the variability of quantitative 2D echocardiographic image texture as a function of range and azimuth must be considered in the clinical interpretation of alterations in image texture. Thus, regional alterations in image texture will not necessarily be due to alterations in the microscopic structure of the tissue being examined. There are several avenues of investigation into the causes and analysis of B mode image texture currently being explored, which may lead to a better understanding of this complex phenomenon. This understanding, coupled with the promising clinical and experimental observations made so far, may make quantitative texture analysis a promising technique for ultrasonic tissue characterization in 2D echocardiography.

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