The Contrast Effect in Ultrasound Image

A. A-S.Mohammed,. I. R-Agool. S. H-Alwan Al-sadi.
Department of laser , Technology university ,
Department of physics College of science Al-Mustansiriyah university,
Department of radiology college of technical medicine.

Abstract

The ultrasonic techniques that are used routinely in several field such as a medical imaging have resulted from basic scientific discoveries, new methods of signal analysis and image processing, the development of transducer materials and fabrication techniques, and the application of digital electronics. The system designer is constrained by the ultrasonic properties of tissues, especially speed and attenuation; these properties determine the optimum choices of ultrasonic frequency, and spatial contrast and temporal resolutions. A new method of detection of defects located in a noisy medium is presented. It is based on considering the ultrasonic grey-level image from the beginning of the analysis. The data processing used here is searching for a certain determinism in the spatial and temporal evolution of the image in the presence of a defect.

Introduction

In any imaging method, the quality of the image is influenced by the spatial, contrast and temporal resolutions of the imaging system, the noise and the presence of artifacts. In pulse-echo ultrasonic imaging, the system designer has to begin by specifying the dimensions of the tissue slice (or volume) that is to be examined. Than, the speed of ultrasound in tissue (about 1500 ms^{-1}) determines the time necessary to obtain a single line of pulse-echo image information (1).

The attenuation of ultrasound in soft tissue is around 0.5$dBcm^{-1}MHz^{-1}$, ranging to about 0.15 $dBCm^{-1}$ MHz^{-1} in fat (2). Generally, ultrasonic imaging systems incorporate a circuit that increases the gain of the receiver with time after the transmission of the ultrasonic pulse (swept gain control) to provide some degree of compensation for attenuation. Swept gain compensation cannot be perfect because of variation of attenuation in different tissues, but contemporary scanning systems have such a wide dynamic range that this is seldom a problem in qualitative imaging (3).

Real tissue are inhomogeneous both in speed and attenuation, and some tissues are significantly anisotropic(4). Quantitative measurement of the effect of tissue inhomogeneity on the shape of an ultrasonic beam is not a simple matter. Typically, abeam may be deviated by 5mm over a tissue distance of 5cm. An alternative, and generally more helpful, approach is to measure the root-mean-square phase aberration due to tissue inhomogeneity across the receiving aperture; typically, this is around 30 ns for an aperture of 15mm and tissue path length of 5cm (5).

Target delectability, which relates to the spatial and contrast resolutions, is further complicated by the image line density and by the presence of speckle and noise in the image texture. Speckle is caused by the phase interference between echoes returning simultaneously from small scatterers within the resolution cell of the imaging system (6). Image speckle patterns do not have a one-to-one correspondence with echo-producing targets in the patient, they seek to improve image quality by traditional methods to produce state-of the art scans of the kind (7).

Image texture analysis is one of many possible approaches to tissue characterization. It depends on a combination of clinical skill and a knowledge of image appearances. So far, however, none of the quantitative analytical methods has turned out to be sufficiently reliable or useful to merit its provision in commercially available scanners (8).

Contrast Agents: A contrast agent is a material that, following administration to the patient, can selectively change the appearance of an ultrasonic scan and thus enhance the ability to detect abnormalities and differences between structures. Particulate suspension and liquid contrast agents have also been tried (2), and quite promising results are now being obtained. If the remaining problems of selective targeting and lack of toxicity can be solved, ultrasonic imaging may inexpensively and safely supplant some contemporary radionuclide and magnetic resonance imaging studies.

We have tried and considered, as a first criterion, the spatial stability of the maximum temporal gradient of each image signal (first hypothesis), as the human eye is attracted by the regularity of the transitions of an image grey-level. As a second criterion, we have opted for the temporal distance existing between the maximum and minimum of each image signal (second hypothesis), as it is known, in this type of testing, a defect will create on the D-scan image a temporally concentrated object.

Detection Algorithm: Let S(i,j) be the original ultrasonic image formed by a set of N numerical signals S_i (t_j) vertically juxtaposed.
This image is intentionally simple to present the algorithm, the performance of which will be displayed later. The image can be written under the following matrix form

$$S(i,j) = \bigcup_{i=1}^{N} S_i(t_j) \qquad \text{for j=1 to M} \qquad [1]$$

in which M equals the number of the image lines. By adiscrete convolution S(i,j) with a vertical gradient G_v, the derived image S` (i , j) is obtained so that

$$G_v = \begin{bmatrix} -1 \\ 0 \\ 1 \end{bmatrix} \qquad [2]$$

S` (i,j) = S (i,j) * Gv [3]
With * denoting convolution operator .
Let it be noted that the derived image S`(i,j) obtained has been formed by the juxtaposition of the derivatives of each original ultrasonic signal $S_i(t_j)$ and consequently can also be written as follows:

$$S`(i,j) = \bigcup_{i=1}^{N} S_i`(t_j) = [S_{ij}`] \qquad \text{For j=2 to M-1} \qquad [4]$$

with

$$S_i'(t_j) = S_i(t_j+1) - S_i(t_j-1)$$

The gradients calculate are positive for increasing variations of the amplitude.

Next, the maxima and minima of the image S`(i, j) are sought column after column. Let G(i) be the maximum amplitude of column (i) and g(i) its minimum amplitude, and G_i and g_i their vertical (or temporal) positions. Thus , we get, for

j = 2 to M-1

$G_{(i)} = S`(i,G_i)$

And

$g_{(i)} = S`(i,g_i)$

A binary image $S''(i,j) = [S''ij]$ is then obtained, in which only the positions of the maxima and minima are considered.

Thus, for each column (i) we have $S''(i,j) = [S''ij]$

And

$S''_{ij} = 1$ if $j=G_i$ or $j=g_i$

$S''_{ij} = 0$ anywhere else

Figure (1) represent the gradient image s`(i,j) calculated from image s(i,j).

Hypothesis: We study the evalution of the position of the max. or min. gradients in image $S''(i,j)$ when the transducer is shifted in acourse parallel to the tissue, i.e in the horizontal direction of the image.

Consequently, a sound stability of the gradient maxima (or minima) will reveal, if the hypothesis put forward proves to be correct, the presence of a defect in the sample under control. The measurement of this stability of position is performed by means of a sliding variance calculation on the binary image $S''(i,j)$. The calculation window F, within which the variance is calculated, is 2f+1 wide and as high as the size of image $S''(i,j)$, i.e. (M-1). It is centered on column (i) , Figure (2)

Optimization of Window F: In an image representative of a structure noise zone, the influence of (f) on the gradient stability is as follows:

- if (f) is too small, the variance of the position of the gradients is unstable. There is a possibility of having a stable event born of the structure which gives a low value of the variance, a would a possible defect. This could lead to a false call.

- if (f) is too large, the variance of the position of the gradients is high but stable. The stability provided by a possible defect cannot be detected, except when the defect size-width (f) ratio is high enough. The detection of small defect is thus impossible.

The selected optimization principle therefore consists of evaluating, in an image of width (L) containing structure noise only, the minimum size of the elementary window f_{opt} for which a second-order stationary state of the gradient position appears. In this way we takes up the detection optimal resolution.

The representation of this trace shows is given in Fig. (3).The operator whether the selected image is representative of the noise, if so, the variance of the variance soon cancels out and the optimal size $2f_{opt}+1$ of window F can be chosen. If not, if the selected image is not representative of the noise and, accordingly, contains a defect-the curve shows a slow decrease, and a new image has to be selected. This study must take place before testing any new sample.

Results

The measurement are performed in immersion, by displacing a transducer (5MHZ / 0.5$''$) in focus (2$''$) step by step (0.1mm and generating transverse waves at (45°). The signals are sampled on 256 points, and the amplitude is digitized on 8 bits.

We dispose of a sequence of images of 200 to 500 signals each, that are shifted square to the tissue. For each image, we calculate the echo dynamics curve. We define the spatial signal noise ratio (SSNR) as the ratio between the max. Amplitudes of the echodynamic curve respectively for the defect and for the noise. In this case, the temporal and spatial signal-noise ratios (SNR and SSNR) can drop below 6dB, making any immediate visual analysis impossible for the operator to carry out.

The testing of the detection algorithm has been performed on (9) defects, differing from one another in –shape , Size , Position.

Fig.(4) shows that analysis, with the first criterion of detection, there is a false call if we use only the first criterion.

The second detection criterion clearly shows a measure of the temporal distance existing between the positions of the max. and min.

of each image signal. That is lower than in the structural noise, fig.(5), the representation, after inversion, shows a peak.

This method of detection determines the sizes of the defects, with a good precision.

Conclusions

This algorithm analyses the ultrasonic image according to two criteria, each of which is good enough alone to ensure the detection of certain defects.

We can note that the defects, in the direction parallel to the image, are defined with a good accuracy.

Finally, this method can be included in an image processing system dedicated to ultrasonic NDT. It will give help to the qualified to perform pertinent defect detection.

Reference

1. Smith, B.J. (1986)" A New Concept in Ultrasonic Instrumentation" J.NDT, pp9.
2. Burch, S.F. (1984) Digital Image Processingin Non Destructive Testing" Research Techniques in NDT vol.7, Academic press, London, .
3. Corneloup, G. and Cornu, B.(1987) " Automatic ultrasonic image analysis for defection " , ultrasonic, October,1057 .
4. Dines, K.A. and Goss, S.A. (1987)" Computed ultrasonic reflection tomography", ultrasonic, pp.309, .
5. Lerner, R.M. and Huang, S.R. (1990) " Son elasticity" images derived from ultrasound signals in mechanically vibrated tissue " Ultrasound Med.Boil, pp231, .
6. Duck, F. A. (1990)" physical properties of Tissue " Academic press London.
7. Tiwells, P. N. (1989) " Dopper Ultrasound in Medical Diagnosis" Br J. Radiol.
8. Thomas, J.G. and Jeanty, k P. (1991)" Automatic segmentation of ultrasound Image using Morpho;ogical operators " , IEEE Trans . Med . Image pp.108,.

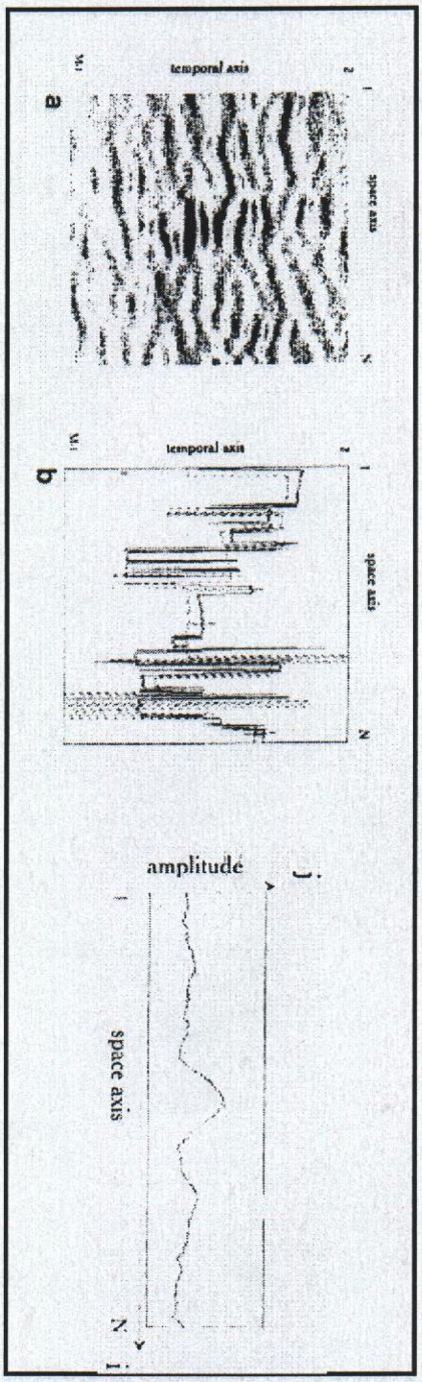

Fig. (1) Two processed image calculated from the original image .

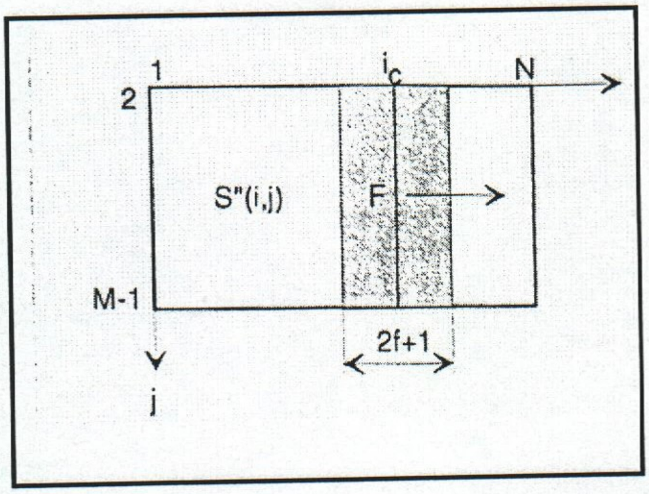

Fig. (2) Definition of the processing window F .

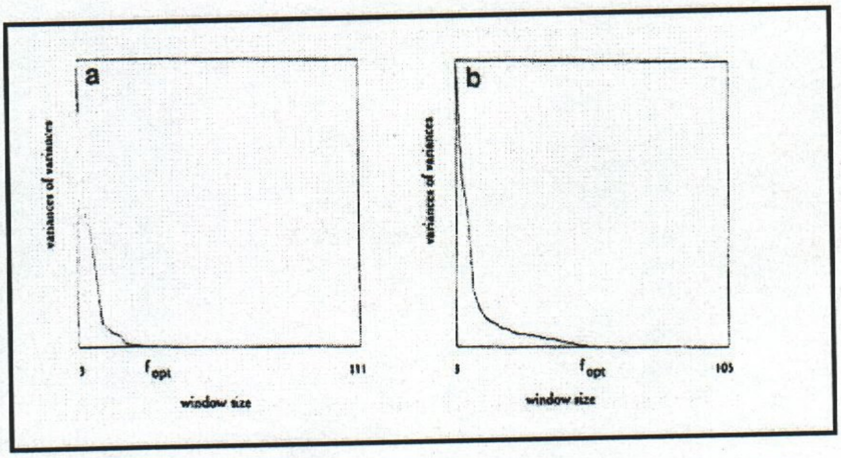

Fig. (3) Calculation of the optimal size of the window F which the curve VV obtained in two cases .

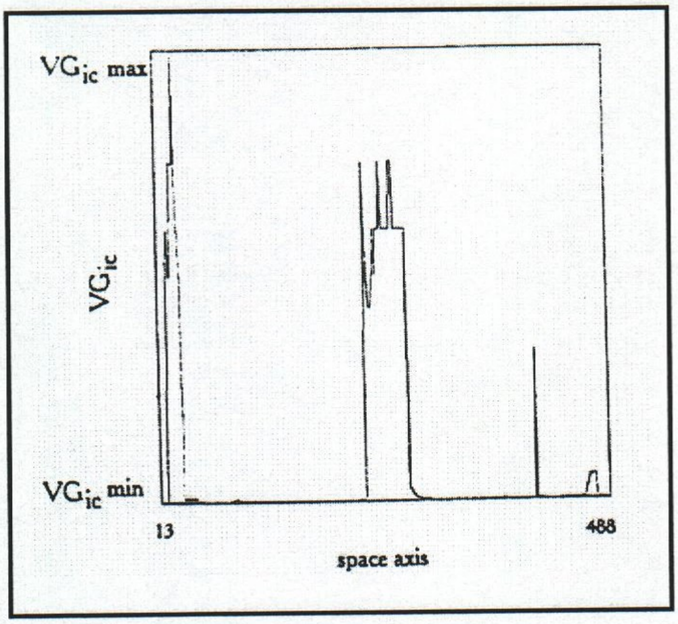

Fig. (4) Curve VG_{ic} obtained on the image

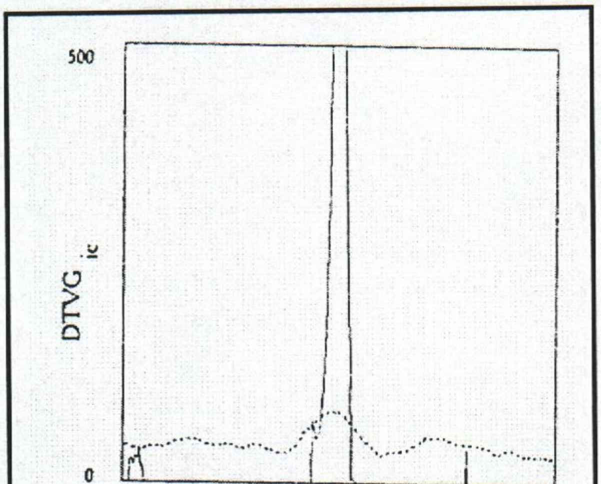

Fig. (5) Curve $DTVG_{ic}$ obtained in the image of figure 1

تأثير درجة التباين في صور الموجات فوق السمعية

عبد الرزاق عبد السلام محمد ، ابراهيم رمضان عاكول و ساجد حمزة علوان السعدي
قسم هندسة الليزر، الجامعة التكنولوجية
قسم الفيزياء ، كلية العلوم ، الجامعة المستنصرية
قسم الاشعة ، كلية التقنيات الطبية

الخلاصة

تعد تقنيات الموجات فوق السمعية المستخدم بشكل روتيني في الصور ولمجالات متعددة و منها الطبية، ناتجة من الاكتشافات العلمية الاساسية، والطرائق الجديدة في تحليل الاشارة، والمعالجة الصورية، وتطور المواد الداخلة في صناعة محول الطاقة (المجس)، والتقنيات المبتدعة، وتطبيق علم الالكترونيات الرقمية، ان مصمم المنظومة مقيد بخصائص الموجات فوق السمعية في الانسجة لاسيما السرعة والتوهين ، وهذه الخصائص تحدد الاختيارات الافضل لتردد الموجات فوق السمعية والتباين المكاني والتحليل الزماني . ان الطريقة الجديدة تكشف لنا عن موقع العيوب في الوسط كثير التشويش. لقد اخذت هذه الطريقة بنظر الاعتبار مستوى تدرج صورة الموجات فوق السمعية من بداية التحليل وان بيانات المعالجة المستخدمة هنا تقودنا الى الفحص الدقيق للتطور الزماني والمكاني للصورة بوجود العيب .