Adaptive filtering for reduction of speckle in ultrasonic pulse-echo images

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Current medical ultrasonic scanning instrumentation permits the display of fine image detail (speckle) which does not transfer useful information but degrades the apparent low contrast resolution in the image. An adaptive two-dimensional filter has been developed which uses local features of image texture to recognize and maximally low-pass filter those parts of the image which correspond to fully developed speckle, while substantially preserving information associated with resolved-object structure. A first implementation of the filter is described which uses the ratio of the local variance and the local mean as the speckle recognition feature. Preliminary results of applying this form of display processing to medical ultrasound images are very encouraging; it appears that the visual perception of features such as small discrete structures, subtle fluctuations in mean echo level and changes in image texture may be enhanced relative to that for unprocessed images.

KEYWORDS: ultrasonic imaging, speckle, image processing

Introduction

The fine structure (texture) of ultrasound pulse-echo images has a mottled or granular appearance which has been called speckle because of its similarity to the equivalent optical phenomenon, laser speckle[1]. The phenomenon of speckle results directly from the use of a coherent radiation for imaging and occurs when structure in the object (in this case the body), on a scale too small to be resolved by the imaging system, causes interference to occur between different parts of the wave received from the object region corresponding to a given point in the image. A useful survey of, and qualitative introduction to, this subject has appeared previously in this journal[2].

The statistical properties of ultrasonic speckle have been predicted by calculation from first principles and by computer modelling, by various authors (see, for example, Ref. 3). When the ultrasonic scatterers are random and finely distributed, so that there are many within the resolution cell of the imaging system, the speckle generated is known as 'fully developed speckle'. Such speckle has a mean and modal value determined by the strength of the scattering but all other properties are independent of the object structure and are characteristic of the instrument used to produce the image. Both the amplitude (grey level) probability distribution and the average speckle cell will have a shape determined by the instrument, which may vary within the image.

It is now agreed by many authors that it would be desirable to remove, or reduce, the speckle in pulse-echo images, since its presence degrades the apparent resolution in the image to a point below the diffraction-predicted value and it interferes with the visual assessment of small differences in mean grey level or texture. A number of methods have been suggested for achieving this, most of the methods involving the averaging of multiple images of the same object structure, varying one or more instrument parameters so that the speckle patterns in the images are significantly decorrelated[4]. This reduces the variance of the speckle modulation and thus makes the mean level and object texture easier to perceive. Additionally, it is possible to reduce the variance of the speckle by smoothing the image with a two-dimensional low-pass filter. A simple filter (even one which varies spatially to account for the spatial variation of the speckle characteristics) will, however, also blur the object-related information which one would wish to preserve. To date, post-formation image filtering methods have not been successfully applied to remove speckle from ultrasonic pulse-echo images.

This paper describes a new method by which ultrasonic pulse-echo images may be smoothed to suppress the fully developed speckle (as defined above),

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while substantially preserving the image component corresponding to resolved (or partially resolved) object structure.

Speckle recognition and filtering

The fully developed speckle occurs when many fine randomly distributed scattering sites exist within the resolution cell of the pulse-echo system. Under such conditions the statistical (texture) features of the speckle pattern represent a multivariate instrument signature. Deviations from this situation (that is, as resolvable, or partially resolvable, structure is introduced) result in deviations in the statistical properties of the image texture[5]. Such deviations result in an image texture which is no longer entirely characteristic of the imaging system. It should therefore be possible to use these deviations to classify each local region of the image according to how much it resembles the fully developed speckle normally generated by that particular imaging system in that part of the image. This measure of similarity can then be used to control the spatial bandwidth of a smoothing filter of some kind, so that regions of the image which closely resemble the fully developed speckle are replaced by a local mean value and, at the other extreme, regions with properties which are least similar to fully developed speckle are not smoothed. Thus the filter would appear to behave intelligently by recognizing those parts of the image which require maximum smoothing.

Various features are available which may be used to describe local image texture, and numerous possible methods of multivariate discrimination/classification exist to combine them to form a measure of similarity. The use of different numbers and combinations of speckle-recognition features will result in a family of filters operating by this method, each offering a different compromise between effectiveness of speckle suppression, loss of real information and speed of computation. The simplest method is to use one, normalized, feature only. This is the approach we have taken to date and has resulted in the filtering algorithm to be described.

Method

To study the potential utility of such a processing technique a filter has been implemented using one feature only to classify the texture. The measure of similarity is simply a scaled, normalized value of the feature itself. An 'unsharp masking' filter[6] provided a convenient way of allowing the degree of smoothing to be controlled by the local features of image texture. This is defined by

$$\hat{x} = \bar{x} + k(x - \bar{x}) \qquad (1)$$

where $\hat{x}$ is the new (processed) value of a pixel to be computed from the old (unprocessed) value, x, and the local mean of the old values surrounding and including that pixel, $\bar{x}$. The constant, k, is controlled by the measure of similarity used, p, which in this case is the deviation in the ratio of the local variance of grey levels to the local mean, $p = \mathrm{var}(x)/\bar{x}$;

$$k = (gp - \bar{p}_s)/p \qquad (2)$$

where $\bar{p}_s$ is the mean value of p in a region of an image which is considered (by human observation) to consist entirely of fully developed speckle. Ideally this should be part of an image obtained using the same instrument to scan a specially constructed phantom consisting of the randomly dispersed fine distribution of scatterers known to produce fully developed speckle, but it may be a relatively uniform region in part of the image to be processed. The constant, g, is a scaling factor which permits the overall aggressiveness of the smoothing to be adjusted.

It may be seen from (1) that regions of the image which have a value of the similarity coefficient, p, which is close to that of fully developed speckle, $\bar{p}_s$, will receive the maximum amount of smoothing, which is defined by the value of g and the number of pixels over which $\bar{x}$ is computed. Note that this is not necessarily the same number of pixels as used for computing the local values of mean and variance which go to form the value of p. Completely objective criteria on which to base the choice of the number of pixels in each of these cases have not yet been developed and at present the minimum sizes of local window which still give pleasing results are used. The

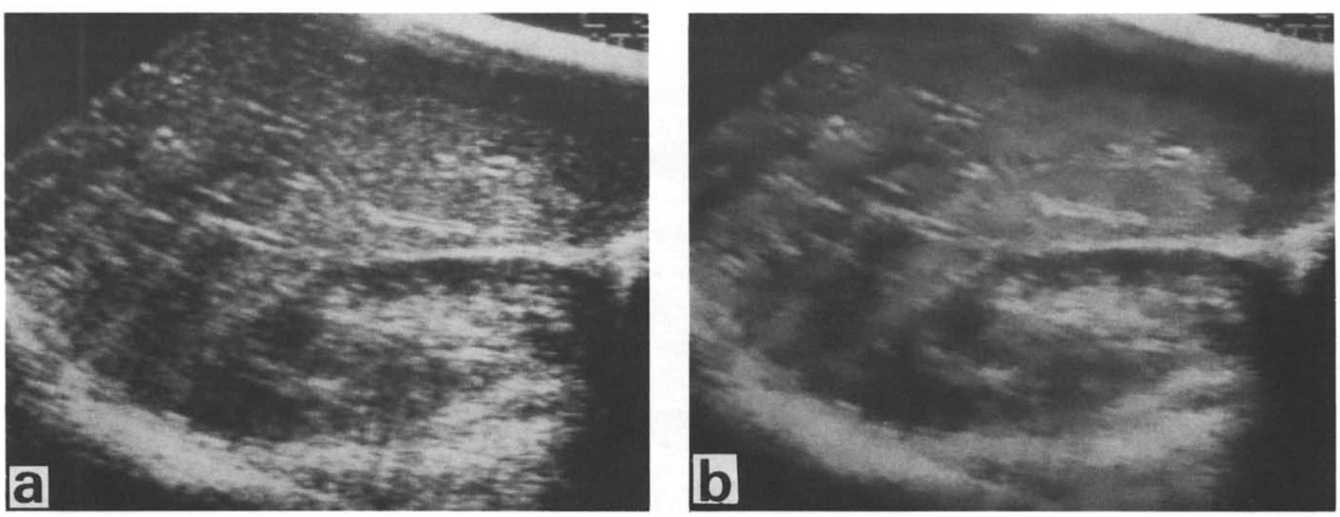

Fig. 1 a— Static (manual) B-scan of a normal liver and kidney in the sagital plane (3.5 MHz, medium focused transducer, image digitized to 256 by 256 pixels by 6 bits). b— Result of suppressing the speckle from the image by using the filter described in the text. A region of the original image (top-centre) was used to provide the reference speckle feature required by the filter

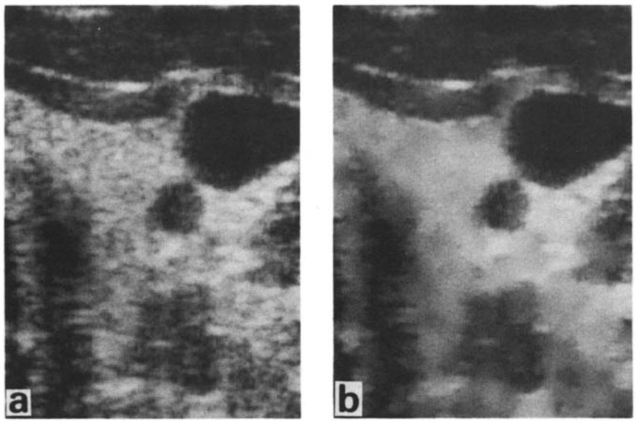

Fig. 2 a — Frozen image from a real-time linear scan of the right lobe of a normal thyroid in transverse section, showing also part of the overlying muscle, the trachea and the carotid artery (5 MHz array, digitization 6 bits and approximately 180 by 200 pixels). b — Image after processing to remove the speckle. The image texture within the thyroid itself was used as the reference speckle pattern in this case

value of the scaling factor g is chosen such that for the image to be processed k is always less than or equal to one. Inspection of (1) will reveal that this constrains the filter to range between doing nothing and maximum smoothing. If k were permitted to become greater than one then some edge enhancement of the image would take place.

Results

Figures 1(b) and 2(b) illustrate the results of applying this processing technique to the two typical ultrasonic images (obtained from different scanners) shown, respectively, in Figs 1(a) and 2(a). In both cases the image region used to provide a reference speckle pattern was found within the image to be processed. Also, a window of 7 by 7 pixels was used both to calculate the local mean and variance and to maximally smooth the image. It is readily apparent from studying these images that the aim of suppressing the speckle while preserving useful information has, in large measure, been achieved. In Fig. 1(b) it can be argued that, relative to Fig. 1(a), it is easier to see the numerous isolated blood vessels within the liver, the relative mean brightness of liver and kidney is easier to assess, and gradual changes in mean brightness within the liver are more obvious (some of these are due to fluctuations in the speed of the manual scan). Similarly, relative to Fig. 2(a), in Fig. 2(b) the subtle fluctuations in mean scattering level within the thyroid, the thyroid boundary and relative brightness level throughout the image are easier to perceive, and the image generally looks much 'cleaner', in the sense of being free from noise.

Discussion

It is of interest to compare our approach with work described in two previous publications. Firstly, this first implementation of our speckle suppression filter is similar to a multiplicative noise filter developed by Lee[7] to smooth synthetic aperture radar images, which also suffer from speckle. Our approach is, however, hopefully capable of considerable development and improvement, based on a multivariate description of speckle, and is not limited simply to filtering multiplicative noise.

Secondly, Dickinson[8] has attempted to smooth speckle in ultrasound pulse-echo images by using an unsharp masking filter, where the equivalent of the coefficient k is controlled by the local mean, $\bar{x}$. His filter was designed to preserve only large amplitude echoes from discrete structures such as blood vessels. It would not, for example, have been able to cope with two different texture regions which possess the same mean level, or with two regions of fully developed speckle which possess different mean levels. For the latter case one would wish to replace the speckle by the mean echo level but preserve the boundary (edge) between the two regions. This is just what happens with our filter, but with that of Dickinson the speckle region of higher mean level would be undersmoothed. The local mean is not a useful feature for speckle characterization. It is in fact the one speckle characteristic which is determined by the object and therefore should be preserved in the final image.

Conclusion

An adaptive two-dimensional filter for the suppression of speckle from ultrasonic pulse-echo images, based on the use of the ratio of the local variance and local mean of the image to recognize those parts of the image which require maximum and minimum smoothing, has been found to perform well in preliminary tests on clinical images. It is expected that the use of appropriate combinations of a range of texture features will produce even better results.

Further developments of this approach to filtering ultrasonic images should also include the use of reference speckle patterns which are spatially variant to allow the filter to deal effectively with the spatial variation of the speckle characteristics of the imaging system. Such reference images could be obtained from phantoms constructed to produce fully developed speckle and to mimic the frequency-dependent attenuation typical of soft tissues. However, the need for such developments may be somewhat offset by two features of current medical ultrasound scanner design. Firstly, in one popular class of scanners the image is produced by sweeping the sound beam through a circular sector. A polar coordinate version of the filter would permit much of the far-field (beam spreading) spatial variation of speckle in sector scans to be automatically accounted for. Secondly, some scanners now incorporate highly sophisticated multiple zone or swept focus systems designed to maintain good lateral resolution over a wide depth range. These systems tend to produce spatially more uniform speckle patterns.

Considerable effort is now required to assess fully the benefits (if any) which this form of image processing might offer to clinical diagnosis by ultrasonic imaging. There may also be advantages to applying the process to B-scan images obtained for other purposes, as in non-destructive testing.

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