Objective Analysis Of Ultrasound Images By Use Of A Computational Observer

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Abstract--**We present a computer-based Computational Observer method for the analysis and evaluation of digitized ultrasound images of the Contrast-Detail phantom, which was developed earlier. This evaluation method evolved from image evaluation studies, which demonstrates that human observer performance is not sufficiently consistent or accurate for objective evaluation of images or imaging system. The Computational Observer measures the detection threshold for imaged targets of known contrast, and computes Contrast-Detail data from this information. We find that our new method meets the criteria of being objective, accurate, reproducible, transportable, and relevant to human observer image evaluations.**

I. INTRODUCTION

The noninvasive nature, low cost, portability, and real-time image formation make ultrasound imaging an essential tool for medical diagnosis. Over the years, its application extended to include many fields and research is underway to improve the technology even further. One of the areas where research in this field has addressed is the fundamental problem of detection of focal lesion, which is a major limitation on image quality in ultrasound imaging[1]. The detection and assessment of focal lesions of low contrast against background tissue within individual organs is an essential task in all medical imaging. Examples of this problem, in diagnostic ultrasound, are the detection of breast mass (cysts or tumors), focal lesions in the liver, or infarcted regions of myocardium. This task is in turn affected by the quality of images produced by an imaging system. Degradation in image quality can affected patient scans and therefore diagnostic may not correctly. Early detection of image quality defects can verify that equipment is operating correctly and repairs are done properly. In the past, many testing methods have been developed for the subjective evaluation of medical image quality. Unfortunately ,most subjective data tends to be nonreproducible, since the sensitivity levels of human observers and their reproducibility depends on many conditions which are unrelated to objective image quality. Factors such as physiological or ambient condition, training, experience, number of tests, etc., can influence the outcome of a human observer study[2]-[4]. Moreover, most human observer analysis are relatively lengthy and laborious. For example, when human observer are used to perform a receiver operating characteristic (ROC) study, the results are more reproducible and accurate than world be the case for other types of tests. However, a proper ROC study for a set of images requires several observer, and can take several week or months of data collection and analysis to complete [5]-[8]. Our desire to produce an objective, quantitative analysis of ultrasound images brought about the invention and use of the contrast-detail (C/D) ultrasound phantom [9] (fig,1), and methods for the evaluation of ultrasound images by use of this phantom[10],[11]. However, the use of human observer for C/D analysis yields a high degree of nonrandom error, and the levels of sensitivity among observers differ widely[3]. It is for this reason that we developed the computational observer (CO) method, which is a quantitative, computer-based process for the analysis of digitized ultrasound image of the C/D phantom. It is anticipated that in the near future, most ultrasound instruments will incorporate facilities for performing such image analysis.

II. RATIONALE

A .Description of the Computational Observer

The computational observer method uses a computer and a set of algorithms, together with an input device (digitizer), to digitize ultrasound images and then uses these images to compute the detectability of test targets, based on a predetermined the criterion, or known algorithm, which discriminates between a visual target and its background. The process takes into account target size, contrast, mean brightness level, noise, and distribution of brightness levels. The CO derives a contrast-detail curve from the computed signal-to-noise ratio (SNR) of the region inside the target, versus the target size, to produce a quantitative estimate of target detectability for a range of target diameters as a function of target size and image contrast. Other investigation have designed computer-based image evaluation system [18],[19], but they are not designed to measure contrast-detail performance.

The CO is based on the most primitive model of detection of visual information in a noisy image. This is known as a level 1 or "known signal in noise" detection problem [12]. The CO's main function is to discriminate a visual target from its background by computing detectability indices for different size target areas, which are then compared to a preselected decision threshold. The CO concentrates on the statistics of the detection process, and makes no attempt to mimic higher level detection, as might be performed by the human visual system.

The design of a computational observer for evaluation of ultrasound image is a difficult problem in comparison with evaluation of other imaging modalities. The difficulty stems from limitations brought about by the characteristics of ultrasound images, the physical properties of ultrasound image noise, and the properties of the current version of the C/D phantom. We point out that ' noise ' as described in this paper, refers to the coherent speckle present in all ultrasound images. Although both electronic and acoustic noise are present in every pixel of the digital ultrasound image, the

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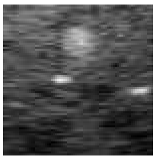
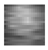


a) Original Image



b) Target ROI

Fig. 1 ROI selection

greatest contribution to the image noise is the speckle, which is different from the type of the noise in photon images, such as X-rays where the noise usually called quantum mottle. Because of the noise content, many spatially uncorrelated images of the same target cross section are required to measure the required detectability parameters, but in our case, due to the limitations of the phantom, we are limited to one sample of a given conical target cross section.

B. Theoretical Detection Criteria

The Rose-DeVries model: The design of the CO is based on the Rose-DeVries detection model from signal detection theory [12], [13]. The criteria in this model, which we describe below, state that the variation in the total number of image quanta account for the detectability of a signal which is just slightly different from its background (i.e, low contrast). Thus, the detectability of a ROI signal in an image is determined by the difference between the number of image quanta contained in the signal region of interest, and those in equally sized samples of the background region of interest in units of noise standard deviation. Furthermore, the ratio obtained from the difference between target quanta and background quanta divided by the noise standard deviation must exceed some number k which is the threshold value of the signal-to-noise ratio (SNR). Rose computed the threshold contrast for many ROI targets of varying sizes, for different human observer data for a limited range of ROI sizes, but was considerably different from measured values for vary large or very small ROI sizes [13]-[15].

C. The Copmutational Observer Performance

The data sampled by the CO are the envelope-detected amplitude signal in the B-mode image over a region of interest (ROI) size which matches the target cross-section area, and a second area, identical in size and shape fig. 1, which is placed over the background at the same depth as the target in the image.

The computational observer sums the brightness amplitude values Vi over all pixels i, in the target area and the background area, respectively, in the digitized ultrasound image. These samples are summed over all pixels, i, for the entire ROI in the digitized B-scan image to produce a mean brightness value. The mean amplitude brightness values <Vi>for each of the many ROI's sampled, are summed to produce an average of the means, μ_i.

The computational observer has access to one image sample, or a limited number of independent image samples of the target, because there exists only one C/D phantom with the properties we have described, and an ultrasound [i]imaging system can produce only one image of a cross-section of a given target. It is possible, however, to estimate the change in variance due to the different-sized ROI's by using many background images, which offer independent ROI's for the calculation of the variance of the average brightness over the background. This gives us an estimate of the standard deviation of the background ROI, but not the target ROI, as a function of area. The standard deviation of the target ROI can only be estimated from the standard deviation of the background ROI's.

Thus, the Computational Observer, using the present version of the C/D phantom, can calculate the standard deviation of the background ROI, or σ_B, as a function of area size, but can only estimate the true σ_T for the target. The CO approximates d' by using the value $(\mu_T - \mu_B / \sigma_B)$. Rose uses the approximation to compute the threshold SNR of between 2 and 5 for human detection of low contrast signals.

The CO repeats the calculation of $d' = (\mu_T - \mu_B / \sigma_B)$ for many different-sized cross-section of a conical target in the C/D phantom, and forms a plot of d' versus target size for the fixed-value contrast of the target. The CO decides whether the amplitude signal has been detected by comparing the computed d' value to the cutoff value, k. If the d' exceeds this value, the signal is considered detected.

III. EQUIPMENT USED IN EXPERIMENT

A. Ultrasound Scanner

Solo Compact ultrasound imaging system (International Biomedical Engineering Technologies, Egypt).

B. Phantom

The C/D phantom Model 84-340 General Purpose Multi-tissue Ultrasound Phantom is constructed from a patented solid elastic material called Zerdine®. Unlike other phantom materials, it is not affected by changes in temperature. It can be subjected to boiling or freezing conditions without sustaining significant damage. It is also more elastic than other materials and allows more pressure to be applied to the scanning surface without subsequent damage to the material. At normal room temperature, Zerdine will accurately simulate the ultrasound characteristics found in human liver tissue. It contains dense and cystic masses in a range of sizes, one high-density target, and an assortment of nylon monofilament target groups. It was designed to allow for assessment of linearity, axial and lateral resolution, depth calibration, dead zone measurement, and registration within two different backgrounds of 0.5 and 0.7 dB/cm/MHz. The phantom is protected by an acrylic case and plastic membrane to facilitate scanning and minimize desiccation.

IV. EXPERIMENTAL METHOD

The Computational Observer process involves the following procedures, summarized from [16].

a) The digitizing of ultrasound B-scan images of many scan planes of the contrast-detail phantom.

b) The sectoring of the digital images, and establishment of location coordinates for each target and background. This may require image enhancement for very low-contrast target.

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c) The summing of pixel brightness values within a ROI, and calculation of the average brightness for that ROI.

d) The repetition of step c) for many identically sized region of the background around the target, and obtaining the variance and standard deviation for the individual average brightnesses.

e) The repetition of steps c) and d) for different-sized regions, corresponding to the different cross-sections of the conical targets in the phantom.

f) The computation of the difference of the average of the means (μ_T - μ_B), divided by the standard deviation of the means for each different cross-section in the conical target, to obtain an index of detectability, d'.

g) The repetition of step f for each different conical target, which corresponds to a different contrast level.

h) The plotting of detectability index versus ROI size for each different conical target (contrast) in the phantom.

i) The finding of a threshold ROI size for each different target by selecting an arbitrary SNR cutoff criterion, and using the plots obtained in step h) to locate the threshold diameter corresponding to chosen criterion.

j) The construction of a C/D plot from the data obtain

in step i) above.

V. RESULTS

A. Computation of Detectability Index

Having calculation the brightness variance inside the ROI, the next step in the Computation Observer process was the computation of the detectability index, based on the approximation $d' \simeq (\mu_T - \mu_B)/\sigma_B$ where μ_T is the mean target

brightness for the given ROI, μ_B is the mean background brightness for the given ROI, and σ_B is the standard deviation of the background brightness for a given size ROI.

This process was repeated for target contrast 0.32, 0.38, 0.51, 0.68 in the phantom, and plots are shown in Figs. 1-4.

B. Construction of C/D diagrams from CO Data

From the data obtained above, Figs.2-5, we can construct a C/D plot. First, we choose a cutoff criterion, d' = k approximate d' by our measured value and we use the criterion for all the different targets for the corresponding threshold square length. We chose, arbitrarily, k = 2.0, (because it is in the expected range of possible criteria for the CO data). (Alternately, we can choose any other value of k, to find a family of C/D curves). We repeated this process for target contrasts 0.32 to 0.68, and constructed the C/D plot shown in Fig. 6.

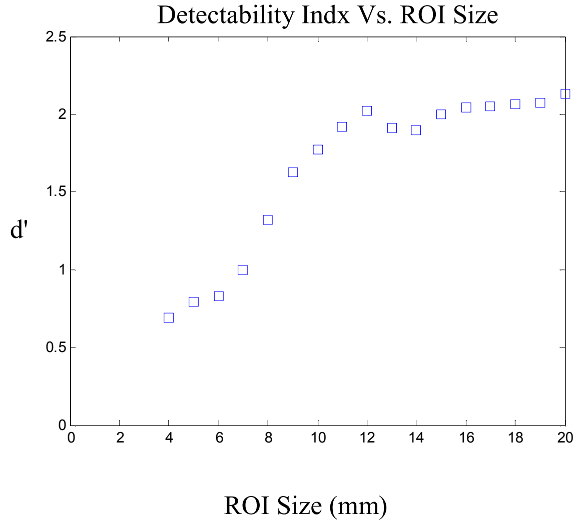

Fig.(2)

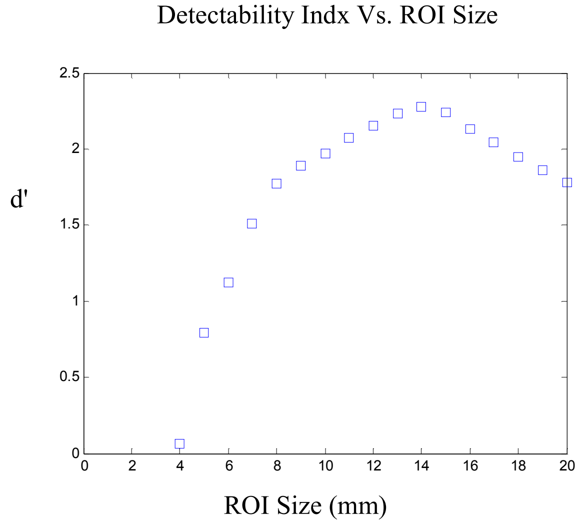

Fig.(3)

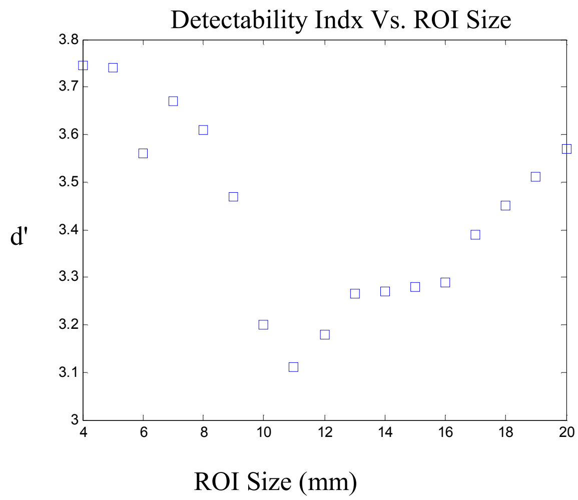

Fig.(4)

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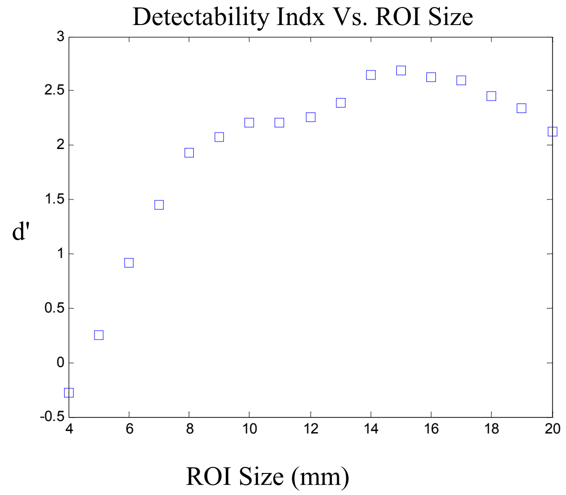

Fig.(5)

C. Error analysis

a) Nonhomogeneous Scatter Concentration at the Target Tips: This problem deals with how the scatterer concentration within a given target may vary from one phantom to another, due to settling or clumping during production. This may give rise to both target contrast errors and visual artifacts.

b)Image Brightness Trend: As we demonstrated earlier, the images used in our experiment were found to contain a subtle, systematic, brightness variation due to attenuation not compensated by the time-gain correction within the image. Since this type of artifacts affects all ultrasound images, it must be corrected or normalized if systems are to be compared to each other. We present a method for image correction and evaluation in [20].

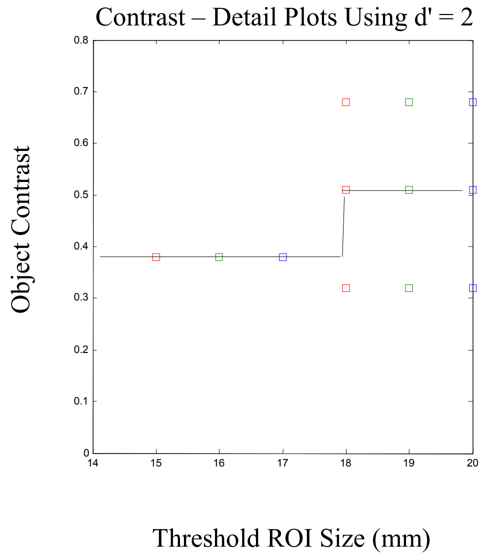

Fig.(6)

VI. DISCUSSION

As we can see on the C/D plot, Fig. (6) for contrast levels of 0.4 corresponding to an ROI size of about 18mm and lower than 0.4 corresponding to lower than 18mm, target ROI size detectability is increased with contrast. Contrast for object (focal lesion) must be high to detect focal lesion against background tissue whether object size small or large while contrast for the entire ultrasound image is lower compared with the contrast for the focal lesion and from Figs. 2-5 can be evaluation of ultrasound images when curve related to normal distribution curve can be decide that ultrasound image is high quality.

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VII. CONCLUSION

We have presented the design, implementation, and testing of a computational observer method for objective evaluation of ultrasound images. This computational method uses digitized ultrasound B-scan images of a test phantom (the contrast-detail phantom), and is able to calculate the detectability of a target (signal) from its background (background noise). This method produces a quantitative detectability index, based on the measured SNR of the image data, and is measured for computational observer.

We have demonstrated that the computational observer method may be a more useful, objective way of evaluation ultrasound images and imaging system, than method that rely solely on human observer. This method may also be applicable to other types (i.e., other than ultrasound) of imaging systems which produce noisy images.

We demonstrate that this method is as follows:

1) quantitative (yield numerical result),
2) reproducible (within a laboratory),
3) absolute (i.e., the threshold used is a predetermined, value which does not depend on the specific laboratory or group of observers),
4) speeds up the evaluation of an image or imaging system (compared to using human observer), given the right conditions and equipment.

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