The Elaboration and Clinical Testing of a New Technique of Image Quality Improvement in Ultrasound Medical Diagnostics

N. S. Kulberg[1], T. V. Yakovleva[1], Yu. R. Kamalov[2], V. A. Sandrikov[2]
L. V. Osipov[3], P. A. Belov[3]

[1] Dorodnicyn Computing Centre of the Russian Academy of Science
Vavilova st., 40, Moscow 119333, RF
[2] National Research Centre of Surgery
Abrikosovsky per. 2, Moscow 119992, RF
[3] Izomed, ltd
Timiryazevskaya st. 1 corp. 2 Moscow 127422, RF

Abstract. The subject of the present research is solving the problem of the noisy and the informative texture elements separation with taking into account the specific traits of the ultrasound visualization. A noise suppression procedure is realized on the basis of the elaborated mathematical model. The elaborated technique has been tested in a clinic. The testing has confirmed its efficiency. The work has been implemented under the support of the Russian Foundation of Basic Research (RFBR), project № 08-01-12011-ofi.

1. Introduction

The main purpose of the investigations in the ultrasound medical visualization consists in getting the most comprehensive information about the structure of the scattering biological tissue on the basis of scattered signal data analysis, and in further localization and characterization of the lesions. This task is far from its final solution. In this field various particular applied tasks appear, such as the resolution improvement, noise influence decrease, etc. Their main point is to insure the capability of the physician to identify the pathological changes on the background of the secondary image elements that appear during the ultrasound image formation due to the influence of various physical factors. The presence of the speckle structure is well known factor in the visualization tasks. The speckle noise inevitably appears in the coherent imaging systems. It is caused by the signals' interference at scattering by the randomly distributed very small objects. The speckle structure is imposed upon the useful texture of the image. This kind of distortions is one of the most difficult for processing.

In the present work the speckle elements origination mechanism has been analyzed. The mathematical model has been developed to describe the basic elements of the speckle structure according to the criteria of clinical usefulness and in correlation with their spectral properties. Image improvement algorithm based upon the proposed model has been developed and tested.

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2. Comparison with Similar Results

Traditionally the noise suppression in digital visualization is based upon *a priori* information about the differences in spatial spectra of the noisy and the useful elements of an image [1]. Thus, for some classes of images (for example, in digital photography) one can introduce a hypothesis that the useful elements' spectrum is shifted to lower band in comparison with the noise spectrum. Images in this case can be improved by low frequency filtering.

Such approach works well only if the useful and noisy elements' spectra can be reliably separated (e.g., for the small-grained photos). Such a processing is not applicable when these spectra can not be distinguished between each other. The coherent images which are characterized by the relatively large-scale interference noise elements cannot be improved by this way: depending upon the filter adjustments either the image remains significantly noised, or the unacceptable meaningful small details loss takes place.

In proposed technique the described drawbacks of the traditional approaches are absent: the sharpness of the boundaries and the small-scale useful elements of the image are kept even at high degrees of noise suppression.

Some investigators undertake the attempts to find such numeral parameter which variations could become an indication of the presence of some pathologies in tissues [2]. These works deal more with the attempts to automate the diagnostics process rather than with the image quality improvement.

The most close prototype of the proposed technique is the method GOPView, that was developed in 1980-s by Swedish company ContextVision [3]. All the software products based upon this method are characterized by a common approach at which the image being processed is described by means of a set of vector "features", the vector direction characterizing the "class" of an object (for example, the direction of the boundary between two areas), and the vector length characterizing the "certainty" of a feature. The image processing by means of this method consists in finding the fields of "features" and their further analysis.

In our approach there is no intermediate step with the finding and analysis of the "features": the initial image elements are analyzed and filtered directly .

3. A Medical Problem Statement

Before discussing the mathematical criteria of the recognition of "useful" and "noisy" elements of an image it is necessary to present a verbal "intuitive" description of the investigated objects in such a way as they are perceived by a physician. Taking into account the medical aspects of the problem together with the formal-mathematical ones allows to increase the chances that the obtained solution will be really useful for medics.

At visual analysis of an image the physician pays attention to two informative levels in the image. They differ by size of the visualized objects relatively to the so-called "resolution spot" of the acoustical system. This term means the minimal image element which is distinguishable by the observer. This size in lateral area is determined by the acoustical system frequency: the smaller the frequency is, the worse its lateral resolution is, and, consequently, the wider the “resolution spot” is. Its longitudinal size is determined by the duration of the irradiated pulse: the longer the pulse is, the worse is the longitudinal resolution and bigger is the “resolution spot”. The easiest way to estimate the size of the “resolution spot” of a certain acoustical system is to observe bright point reflectors of the acoustical phantom.

The first informative level (mentioned above) is a large-scale structure of the object under investigation. At this level one can estimate the boundaries of organs and their mutual disposition, the focal lesions and the large vessels. The macrostructure is determined by the objects which size is larger than the “resolution spot” of the acoustical system. Due to this fact the definition of the correspondence of the macro-objects in the acoustical image to real anatomic objects and the estimation of these objects’ properties is implemented relatively easily (we do not discuss here the various artifacts that can make the task more difficult). At diagnostics process the “direct” features are estimated: the organs’ deformation (local or diffuse increase or decrease, the presence of the local outpouching or retraction) and the violation of their normal topography, the smoothness and clearness of the contours of the focal lesions, the uniformity of the diameter and the right run of vessels.

The second informative level is presented by the small-scale structure. Here we mean the various texture filling of extensive and relatively homogeneous areas of an image of the parenchymatous organs. The structure of the scattering medium in this case is characterized by the presence of a lot of scattering objects whose characteristic size is of the order of tenth and hundredth portions of a millimeter. These are the the cells layers of various orientation, arterioles and venules, small ducts, fibrous and fat interlayers. At disease the morphological structure of these objects is changed. This may lead to the changes of their acoustical properties. The local variations of the brightness caused, for example, by small vessels, single small stones, etc., can also be attributed to the small-scale structure level.

Thus, the microstructure of an image is determined by the scattering objects which are smaller than “the resolution spot” of the acoustical system. An acoustical image of such objects in principle does not have any geometric likeness with the original object whereas within a single “resolution spot” there are tens and hundreds inhomogeneities which are added statistically. As we deal here with the coherent adding of the acoustical oscillations, an interference amplification or weakening of the scattered signal take place, which are not directly caused by the real changes of the reflective characteristics of micro-objects (speckle-noise).

Nevertheless the change of the micro-objects’ scattering properties at various deceases leads (though not directly) to the change of the image features. That is why the characteristics of the texture structure of the picture that is obtained from these micro-objects (including the speckle-noise) can serve as an indirect source of the information about the disease nature. The changes of the brightness (the signal intensity), the texture homogeneity and the degree of the attenuation of ultrasound signal are estimated.

So, the morphological features of the objects under the study at macro- and micro-levels are principally different. This fact causes the difference in approaches to their analyzing. Obviously this leads to choosing the different mathematical means for these informative levels’ processing.

4. Separation of Macro- and Micro- Structures

A two-dimensional discrete function f_{ij}, obtained as a result of ultrasound transducer scanning along one special coordinate x, is being processed. Depending upon a real scanning coordinates system this coordinate may correspond both to polar angle (the convex and the sector transducers) and to one of the Cartesian coordinates (the linear transducers).

An image comes to processing in the coordinate system of scanning, i.e. before passing the scan-converter. These data have already been passed the procedure of quadrature detection and logarithmic compression of the dynamic range.

For convenience of the mathematical estimations let us transfer from discrete functions to the function of continuous variable: $f_{ij} \Rightarrow f\left(x_i, y_j\right) \Rightarrow f(x,y)$. This is possible under Nyquist-Shannon sampling theorem for functions having the finite spectrum. It should be noticed that the latter condition, i.e. the limited spectra's ranges, is valid with some approximations. However its validity is believed to be sufficient for the most part of practical cases.

In order to separate the small-scale and large-scale image structures let us use the procedure similar to the known method of pyramidal coding. We shall represent an initial image $f(x,y)$ as a sum:

$$f(x,y) = M_0(x,y) + m_0(x,y) \tag{1}$$

Here $m_0(x,y)$ is a function of the small-scale structure including mainly the information about small details of the initial image.

The function of large-scale structure is the following:

$$M_0(x,y) = f(x,y) * \psi(x,y) \tag{2}$$

It contains the information about large details of an initial image and is calculated by means of the low frequency filtering (LFF) of the initial image function $f(x,y)$. Here $\psi(x,y)$ is a pulse characteristic of some LFF, a sign $*$ means convolution. Let us denote the spatial spectrum of this filter as $\Psi\left(\Omega_x, \Omega_y\right)$.

As the simulation results have shown the type of the LFF is not critical. The only critical condition is the velocity of the monotonous decrease of the filter amplitude-frequency characteristic (AFC) behind the cut frequency, which must be not less than 10 dB per octave. The principle meaningful factor is the choice of the cut frequencies, what directly depend on the specific traits of the acoustical image acquisition. Let us consider this issue in more detail.

The most important parameter of the images under our study is the so-called local correlation radius. The results of many theoretical and experimental investigations have shown its value to be determined mainly by the ratio between the emitted signal's wavelength, the ultrasound pulse duration and correlation characteristics of the object under the study. For the homogeneous areas which are characterized only by the presence of a texture, the correlation radius is the least one and is determined by the sizes of the "resolution spot" of the acoustical transducer R_C. The filter cut frequency is determined by the following formula:

$$\Omega_C = 1/R_C \tag{3}$$

The two-dimensional filter $\psi(x, y)$ can be built as a superposition of one-dimensional LFFs with the cut frequencies calculated by the formula (3).

It should be noticed that the requirement to the scale properties of the filters to be used depending upon the correlation characteristics of an image makes it difficult to apply the wavelet transform, at which the spectral filter characteristics are rather dependent upon the scale of the discrete image.

To build an improved image $f_E(x, y)$ we shall use an auxiliary "reconstruction function" $R(x, y)$ for macrostructure and $r(x, y)$ for microstructure:

$$f_E(x, y) = M_1(x, y) + m_1(x, y) = R(x, y)M_0(x, y) + r(x, y)m_0(x, y) \tag{4}$$

The key moment of the present work determining its scientific novelty are the methods of functions $R(x, y)$ and $r(x, y)$ calculation. These functions represent the numerical criteria of the "usefulness" of any image element. Determining the function $r(x, y)$ is of particular importance for us as the analysis and the processing of the macrostructure, according to the authors' opinion, present a less complicated task than the processing of the textured "homogeneous" areas.

5. The Macrostructure and the Microstructure Processing

The macrostructure processing is relatively simple to implement. The main objects of interest in the images determined by the function $M_0(x, y)$, are the large objects' boundaries. That is why the processing and improvement of such images can be considered as a kind of automated "retouch". These techniques are well known and widely used in various graphical applications. For example, the most simple and evident is the usage of an un-sharp masking method with the threshold criteria [1]. The other ways of processing are also possible and their elaboration is not a difficult task. That is why we shall not consider them here in detail. Significantly more difficult and interesting task is the microstructure processing.

In order to explain the essence of the technique of the small-scale structure processing developed in the present work, let us describe its main components and provide their mathematical grounding. We shall suppose that the signal is emitted as the focused Gaussian beam and that the signal spectrum is also Gaussian. We shall consider the correlation radiuses R_C for both coordinates to be the same (this condition is easy to implement by simple change of one of the coordinates).

The main elements of speckle-structure are the "interference maximum" and the "interference zero" (Fig. 1). Besides, some components of the macrostructure persist in a function $m_0(x,y)$: first of all, the brightness “jumps”, determining the sharpness of the objects’ boundaries.

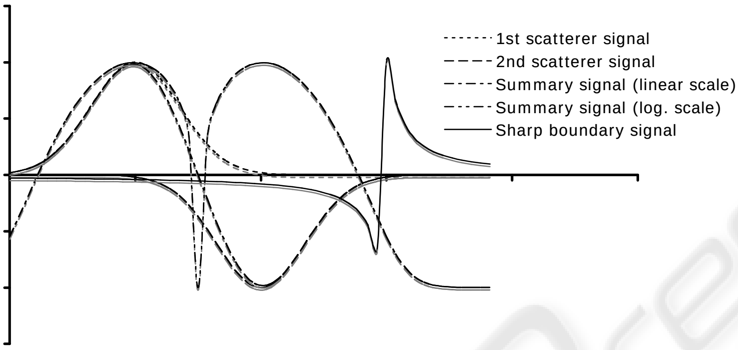


Fig. 1. Basic elements of the small-scale structure represented by the function $m_1(x,y)$ (the cross-sections of the values of two-dimensional function are presented along x-axis). Two antiphase scattering objects result in the oscillation signal having two interference maxima (useful) and one interference zero (harmful)

The interference maximum can be presented approximately as a logarithm of an initial Gaussian “resolution spot” with some limit of the special extension:

$$\left(R_C^2 - y^2\right)\left(R_C^2 - x^2\right)\mathrm{rect}\left(x/R_C\right)\mathrm{rect}\left(y/R_C\right) \tag{5}$$

Having implemented the Fourier transform we obtain the spatial spectrum that decreases while the frequency grows with the velocity $\sim 1/\Omega^2$:

$$\left(\frac{\sin\Omega_y R_C}{\Omega_y^3 R_C^3} - \frac{\cos\Omega_y R_C}{\Omega_y^2 R_C^2}\right)\cdot\left(\frac{\sin\Omega_y R_C}{\Omega_y^3 R_C^3} - \frac{\cos\Omega_y R_C}{\Omega_y^2 R_C^2}\right) \tag{6}$$

Within the surroundings of the “interference zero” the image function’s behavior is, as a rule, linear, before the calculation of its logarithm. After the logarithm calculation we get a function

$$\ln|x|\,\mathrm{rect}\left(x/R_C\right)\ln|y|\,\mathrm{rect}\left(y/R_C\right), \tag{7}$$

with the spectrum :

$$\left(\frac{1}{|\Omega_x|}\frac{1}{|\Omega_y|}\right) * \left(\frac{\sin\Omega_x R_C}{\Omega_x R_C}\frac{\sin\Omega_y R_C}{\Omega_y R_C}\right) \tag{8}$$

At last, the extensive boundary of two objects that is oriented along y-axis, is described in the function $m_0(x,y)$ by the following expression:

$$\vartheta(x) * (\delta(x) - \psi(x, y)), \quad (9)$$

where $\vartheta(x)$ is a Heaviside function. The spectrum of (9) is the following:

$$\frac{1}{\Omega_x} \cdot \delta(\Omega_y) \cdot (1 - \Psi(\Omega_x, \Omega_y)) \quad (10)$$

Having averaged the obtained characteristic for all the angles of the boundary orientation we get the formula:

$$\frac{1}{\Omega_x^2 + \Omega_y^2} \cdot (1 - \Psi(\Omega_x, \Omega_y)) \quad (11)$$

From (11) it follows that the function will have a local maximum near the filter cut frequency $\Psi(\Omega_x, \Omega_y)$. The presence of this maximum can be easily proved taking into account the above introduced condition concerning the function $\Psi(\Omega_x, \Omega_y)$ decreasing velocity.

The usefulness of three described elements is estimated on the basis of the analysis of the investigated data distribution. Traditionally the behavior of the local amplitude of the speckle-noise (let us denote it as v) is described by means of the Rayleigh distribution:

$$F(v) = \frac{v}{v_0^2} \exp\left(\frac{-v^2}{2v_0^2}\right), \; v > 0 \,. \quad (12)$$

However this distribution is not applicable for all the cases (e.g., *see* [4], p. 213). As an evident case of "non-Rayleigh" reflective medium one can propose the model of a medium consisting of a lot of the reflectors of the same amplitude v_0 and random phases, these reflectors being distanced by R_C. The amplitudes of signals that have been received from such a medium obey to the so-called β-distribution:

$$\beta_{\frac{1}{2},\frac{1}{2}}(v) = \frac{v_0}{\pi\sqrt{v(v_0 - v)}}, \; v \in (0, v_0) \quad (13)$$

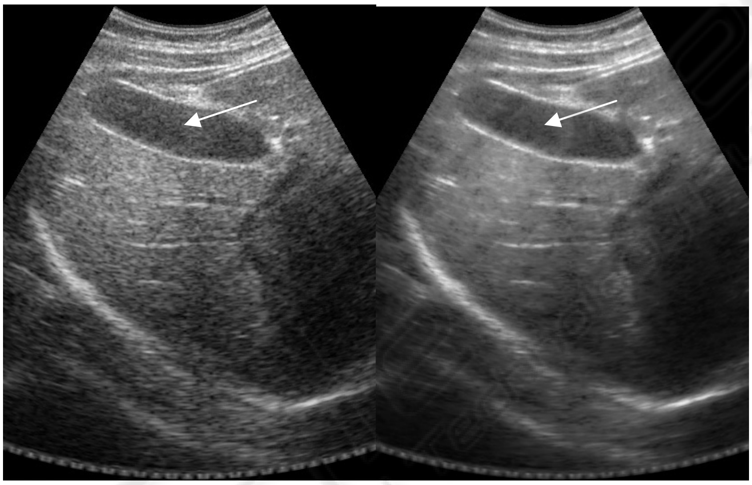
This distribution has a U-shaped density function, while the reflectors' amplitude v_0 which is to be estimated by us, is the upper boundary of the distribution values.

The distributions (12) and (13) just represent two extreme cases of the interference picture behavior, and in practice we have something average. It is important for us that the value under the estimation always lies between the mathematical expectation and the maximum of the real distribution. This fact leads us to the following practically important conclusion: the "**interference zeros" in the image should be believed to be more "harmful" elements of a texture than the interference maxima.**

The comparison of the objects' spectral characteristics with the degree of their usefulness has shown that the most "harmful" object is characterized by slower decreasing of its spatial spectrum, and the most "useful" one possesses the local spectral maximum near the frequency Ω_C. For this reason we propose to suppress harmful texture details and to amplify useful ones by means of a filter characterized by the amplification of the spatial frequencies within the range from 0 up to Ω_C, and by suppression of higher spatial frequencies:

$$\tilde{m}(x,y) = m_0(x,y) * \psi_1(x,y) \tag{14}$$

The certain shape of the filter pulse characteristic $\psi_1(x,y)$ is determined empirically based upon the subjective estimation of the acquired image quality. It depends upon the specific conditions of the measurements.



a) *b)*

Fig 2. An oblique scanning from under the right subcostal area: image of the gall-bladder (marked by arrow) and right lobe of the liver. *a*) unprocessed image; *b*) processed image. After the processing one can notice an improvement of the visualization of the walls of gall-bladder, what makes it possible to get more reliable information about its shape. It is also possible to get a more reliable estimation of the state of the internal gall-bladder's contents

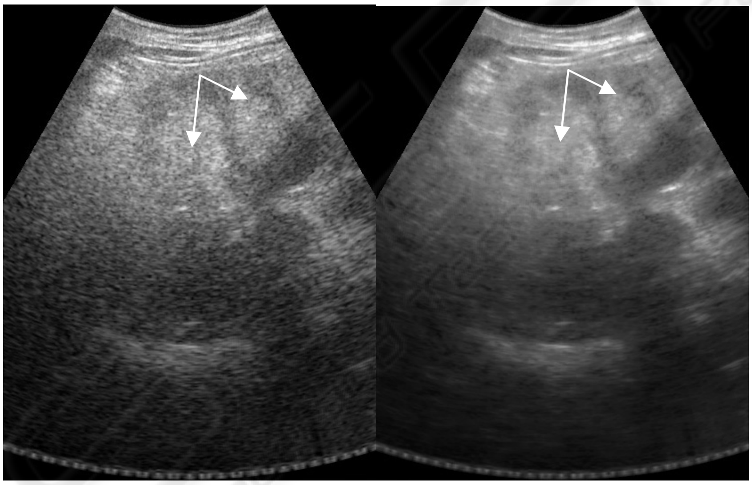
The main property of the function $\tilde{m}(x,y)$ consists in the fact that its **absolute value in any point of the image increases with the degree of the usefulness of an object situated in this point.** Consequently this function can be used as a function of the microstructure reconstruction in formula (4). For additional tuning of the reconstruction function properties one can introduce the nonlinear scaling of the small-scale structure reconstruction function:

$$r(x,y) = \left(\alpha \left|\tilde{m}(x,y)\right|\right)^{\gamma} \tag{15}$$

where the α and γ are parameters of the algorithm being available for the user adjusting. By means of changing the parameters α and γ one can achieve practically a complete suppression of the small-scale texture areas of an image without loss of large objects boundaries' sharpness. The usage of the above described macro-objects "retouch" procedure can still make them more clear. Thus processed picture approximately resembles MR-image and can produce a good impression upon non-specialists. Nevertheless, such excessive noise-suppression, according to the physicians' estimation, is not desirable and is even harmful.

6. Clinical Approbation

The techniques developed in the present work have been implemented as a software package called RASP System®. This software was installed in the medical ultrasonic scanner A-4000 developed by Russian company Izomed, ltd. This scanner has been approbated in the process of ultrasound examinations of patients in National Research Centre of Surgery. Some results of clinical testing are included into the present paper.



a) *b)*

Fig. 3. Scanning of the subcostal area from the right: image of the focal lesion in a liver (the nodes of liver cells carcinoma are marked by arrows) on the background of the liver cirrhoses. a) unprocessed image; b) processed image. The processing makes the contours of the liver lesion more clear due to improving the visualization of the hypo-echoic area around it, and also the visualization of the nearby situated hypo-echoic node on the background of the diffuse changes of the liver parenchyma

During the tests we have compared the results of ultrasound visualization of some organs of health people with the data obtained while examining the patients with various diseases. These approbations have confirmed the efficiency of the proposed techniques of the ultrasound image visual characteristics improvement, and, consequently, a real possibility to increase the informative capacity and reliability of ultrasound medical diagnostics at the detected images processing. The results of this processing are presented in Figs. 2—3.

At visual analysis of various abdominal objects images obtained in normal condition and in the cases of various diseases, by means of our program it has been found out that:

1) The method makes it possible to make more certain conclusion about the shape of the texture objects (liver, pancreas, spleen, kidneys), that provides the opportunity of more confident their differentiation from the surrounding tissues and structures, to get more reliable opinion about the smoothness or non-smoothness of these objects boundaries and more precise definition of their sizes.

2) While using this processing one get the possibility of more assured revealing of the focal lesions of the liver (both hyper-, and hypo-echoic) due to having more accurate information about their boundaries, sizes and internal structure.

3) Clearer contouring of the liquid-containing objects (vessels, bile ducts, cysts) allows more assured differentiation of these organs from the surrounding medium and more reliable judgments about their anatomic construction.

4) The larger range of the contrast if compared with the standard imaging modes, makes it possible to reveal more reliably the sludge (sand, small stones, etc.) in bile ducts, and also to differentiate the liver ligaments.

7. Conclusion. The Perspectives of Further Investigation

The present paper is devoted to the processing of significantly “roughed” signals that have passed the quadrature detection and the logarithmic compression of the dynamic range. This results from the obvious fact: in most modern series devices it is relatively easy the get an access to such a data. That is why, in spite of the obvious insufficiency of these data their investigation and processing continue to be of a significant interest. However according to the authors’ opinion the most perspective direction in the ultrasound images improvement is the investigation and processing of the radiofrequency “raw” data which are not practically used at the present time. Receiving the “raw” data in series devices will significantly extend the boundaries of the ultrasound visualization due to the introduction of new techniques of the digital signal processing.

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