Novel Method of the Noise-Reduction in 3D X-Ray Computed Tomography

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

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

Abstract. There is described a novel technique of the noise suppression in X-Ray computed tomography (CT) images. At describing this technique a special attention is paid to the subsequent 3D reconstruction of the human body internal structures being investigated. The work has been implemented under the support of the Russian Foundation of Basic Research (RFBR), project № 08-01-12011-ofi.

1 Introduction

The necessity of the noise reduction in CT images is caused by the fact that the contrast of the soft tissues X-ray images is rather small (congruous to the noise level or even less than noise), so that it does not allow separating the useful elements of an image from the noise. By means of the visual analysis of two-dimensional picture a qualified specialist is able to see and to identify the details of interest even in significantly noised image. However in many practical cases it has appeared to be impossible to acquire the accurate 3D images of the needed structures by means of the software algorithm being used for the volume image reconstruction. At the same time the 3D computed tomography reconstruction of images is an important (in some cases necessary) diagnostic tool. That's why the expanding of the scope of its application is rather an actual task. The principle advantages of 3D reconstruction are as follows:

1) The possibility of the detailed examination of an object of interest in various planes and at various angles;

2) The possibility of making measurements of the volumes of objects having the complicated geometrical shape;

3) The possibility of getting a more clear and demonstrative representation of the specific interconnections of various anatomic formations and structures.

These advantages of 3D reconstruction increase the diagnostic informative capacity at the examination by means of multi-slice spiral computed tomography (MSCT), enhance the operator's assurance in diagnose conclusion and, besides, provide the non-specialists in ray diagnostics the possibility of evaluating the results of the pro-

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posed diagnostic method rather clearly what in its turn allows determining the proper tactics of the patients' cure.

The diagnostics informative capacity of 3D reconstruction of the computed tomography images in its many aspects is caused by the quality of initial images, especially by the ratio of the useful information to the noise, what is of special importance at contrasting the objects of interest for their further more detailed analysis.

2 Comparison with Similar Techniques

The principle point of the traditional and mostly investigated techniques of the digital images noise suppression consists in frequency and statistical filtering [1]. However the techniques based upon just a simple filtration work mainly without taking into account the differences in geometrical properties between the useful and the noise elements of an image. That is why these techniques have appeared to be of low efficiency and they are not widely used in practice.

There is a significant number of papers which investigate the approaches based upon the threshold processing of wavelet-transformation of an image [2]. The further development of the idea of wavelet-transformation has lead to the so-called curvelet transform that implies the basis decomposition functions to be determined to not only a scale of the image elements but also to their angular attitude [3].

The most well-known commercial product in the field of the image improvement techniques is the GOPView technology that has been elaborated by Swedish company ContextVision [4]. Besides of other fields of application this technology is used for the purposes of computed tomography images improvement. The operation of this technique is characterized by a common approach at which the image under investigation is described by means of a set of vector "features", so that the vector's direction characterizes the "class" of an object (for example, the orientation of the boundary between two areas) and the vector's length characterizes the so-called "certainty" of the feature's revealing. In this technique an image processing is implemented by means of the determination of the "features" fields and by their consequent analysis.

In an approach that is proposed by us in the present paper these is no need in an intermediate step of the above-mentioned technique. Namely, in our approach the step of the "features" separation and analysis is absent: just the elements of the initial image are directly subjected to the analysis and filtration. The so-called «transparency masks», being used in our technique are well adapted to the image elements with various angular attitude. This is a distinctive feature of our technique if compared with the wavelet-filtration. At the same time the proposed technique is significantly simpler from the viewpoint of the computational complexity than the curvelet transform.

3 Brief Description of the Noise Suppression Technique

In the present work we apply the method that was previously elaborated by us for the acoustic images noise suppression. The applicability of such an approach to images of different physical nature will be considered below. Let us describe here the principles of the processing algorithm operation. The papers [5,6] present a detailed description of this algorithm.

An initial image $f(x,y)$ can be presented as a following sum:

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

Here $m_0(x,y)$ is a small-scaled image structure including mainly the information about small elements of an initial image.

The function of the large-scaled structure

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

contains the information about rather large elements of an initial image and can be calculated by means of the application of a low frequency filter (LFF) to the initial image function $f(x,y)$. Here the function $\psi(x,y)$ is a pulse characteristic of the LFF, and the sign $*$ means convolution.

For the construction of improved image we use the auxiliary "reconstruction function" $R(x,y)$ for macro-structure and $r(x,y)$ for micro-structure:

$$\begin{aligned} f_E(x,y) &= M_1(x,y) + m_1(x,y) = \\ &= R(x,y)M_0(x,y) + r(x,y)m_0(x,y) \end{aligned} \tag{3}$$

The functions $R(x,y)$ and $r(x,y)$ represent the numerical criteria of the "usefulness" of any element of the image.

To determine the function $R(x,y)$ a simple threshold algorithm is applied what allows increasing the brightness of the output image at the boundaries of the extended bright areas and near the points of significant brightness jumps.

The function $r(x,y)$ is determined by the following formula

$$r(x,y) = \left|\alpha m_0(x,y) * \psi_1(x,y)\right|^{\gamma} \tag{4}$$

where $\psi_1(x,y)$ represents a pulse characteristic of the LFF and the parameters α and γ are the user-tuned parameters of the algorithm.

This algorithm was elaborated for the ultrasound diagnostics images speckle-noise suppression. At developing the mathematical foundation of this algorithm we have proved the spatial spectrum of such a noise decreases with the frequency increase in average inversely proportional to the frequency $\sim 1/\Omega$, while the spatial spectrum of the useful elements of an image is characterized by the decrease velocity $\sim 1/\Omega^2$. This difference between the spectrum decrease velocities for various image elements is the main point of the "transparency masks" construction algorithm for the image texture noise suppression.

4 The Applicability of the Method for CT Images Processing

First of all, one should consider if the approaches having been developed for the ultrasound images processing are applicable to the processing of computed tomography images or not, as the physical principles of these images construction are absolutely different.

The common principles of the X-ray computed tomography are well known [7], and there is no need to describe them here in detail. Let us describe here only the aspects that are of principle importance for the operation of our algorithm.

The ultrasound speckle-noise is determined by he interference of the waves of rather big wavelength (of the order of sub-millimeters) and, consequently, its spatial correlation radius is comparable with the wavelength.

The nature of the noises in computed tomography image is quite different. Such an image is obtained by means of the digital processing of a lot of transmission X-ray shots. There are two principle mechanisms of the noise appearance in an X-ray image: the fluctuation of X-ray quanta number that is registered by a receiver surface unit (the so-called quantum noise), and also the fluctuations caused by the characteristics of the digital receiver elements. In digital systems both mentioned noise components have a zero radius of the spatial correlation, and, consequently, are characterized by the non-decreasing spectrum (the so-called “white noise”). There is a third component of the noise caused by the influence of the secondary irradiation, but we shall not consider it here, as it is negligibly small in modern X-rays systems.

The number of the registered by the receiver quanta is obeyed to the Poisson’s distribution. In practice due to the central limit theorem we get a normal distribution with the average value and the dispersion proportional to the value $I_0\mu$, where I_0 is the intensity of the initial irradiation, and μ is the average linear coefficient of the irradiation attenuation in a medium.

At formation a tomography image a set of transmission pictures is subjected to the Radon transform. As this transformation is linear, both above mentioned estimations remain valid: the tomography image noise remains “white” and obeys to the normal distribution.

As the formula (4) is derived by us just based upon the fact, that the noise spectrum decreases slower than the image useful element spectrum, it is evident that this formula for the mask of the small-scaled structure reconstruction is applicable in the case of computed tomography images as well. On the other hand, as the noise obeys to the normal distribution, one can confirm that for the extended homogeneous areas of an image the function $M_0(x, y)$ gives the best estimation of the sampled average value.

At the present time the separate two-dimensional slices of a tomography images are subjected to the processing (although in a perspective the possibility to expand this method for a three-dimensional case is also considered). The read-offs of an initial tomogram come to the processing in Hounsfield units (HU), which are derived from the linear absorption coefficient μ by the following expression:

$$H(x, y) = 1000 \times \frac{\mu(x, y) - \mu_w}{\mu_w}$$

where μ_w is the coefficient of the irradiation absorption in a distilled water. The proposed algorithm of the calculation of a function $R(x, y)$ does not work correctly with the initial values H , that's a reverse transformation of these values to the values of the linear coefficient μ has to be implemented.

5 The Conditions of the Clinical Measurements

In the process of the computed tomography investigations implementation a series of strictly spatially oriented (according to the assigned parameters) 2D images sets is obtained.

The initial data (the series of 2D CT images) are obtained at multi-slice spiral computed tomograph (MSCT) Somatom V2 of Siemens company (Germany) at usage of a standard three-phased protocol of the liver examination.

This protocol consisted of : 1) native phase (without contrasting); 2) arterial phase and 3) venous (portal) phase with contrasting. For obtaining the arterial and venous phases 100 ml of contrast solution (Omnipak 350 and Vizipak 320) were introduced via bolus injection into the ulnar vein through the 18-20 G catheter.

The parameters of computed tomography images for the phases of contrasting were the same: 160—200 mA (depending upon the patients constitution), 120 kV, the vision field of 30—40 cm, collimation of 1,5 mm and 3 mm, the table velocity: 12,5 mm/rotation, the rotation velocity: 0,5 sec. The reconstructions were implemented every 0,8 mm for arterial phase, and every 1,5 mm for venous phase. The average density in the liver vessels after contrasting was at least by 30 HU higher than the average density of the liver parenchyma.

By using the knowledge about the disposition of 2D images in 3D space and applying the special algorithm a 3D array of computed tomography data has been obtained. An analysis of the series of CT images was implemented at the workstation MultiVox 2D/3D, realized at a personal computer within the compound of an automated radiology system (ARIS MULTIVOX, Russia). The techniques developed in the present work have been implemented as a software package called RASP System®. The image processing program and a detailed report on the clinical trials is available at http://www.rasp.su.

Depending upon the diagnostics purposes for each of three phases of the examination protocol the following modes of the 3D visualization were used:

1) The multi-planar visualization with the choice of an arbitrary cut or without it;

2) Segmentation of the objects of the interest based on the density thresholds ("automatic" segmentation) or manual segmentation of the objects of interest. For MultiVox 2D/3D at the segmentation of the liver vessels the method of their extraction from 3D data array by the threshold density with the subsequent manual linking or the objects having the similar density and being distanced from each other by 1 voxel;

3) The volume visualization with or without minimal or maximum intensity projections;

4) The combination of various modes, the usage of various of "transparency" conditions.

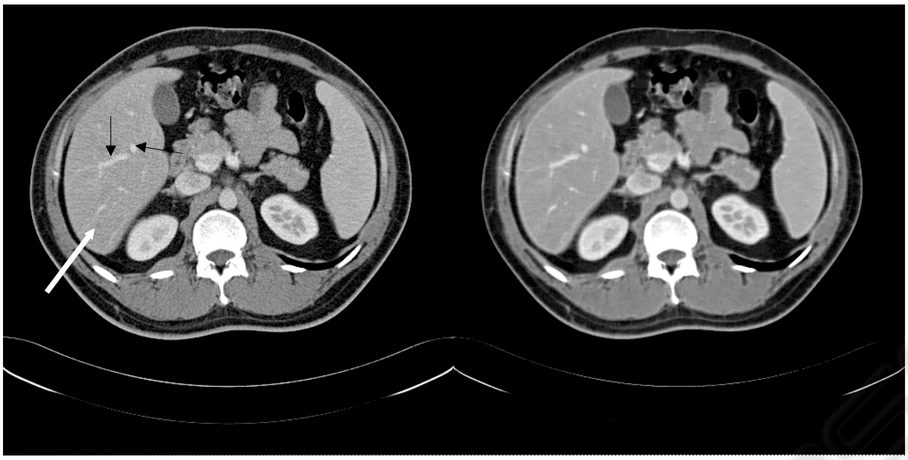

Fig.1. Two-dimensional slices of tomography images before and after the processing at an average level of noising the picture. Liver (thick white arrow), contrasted vessels (thin black arrows). The principle attention was paid to the improvement of the contrasted liver vessels visualization.

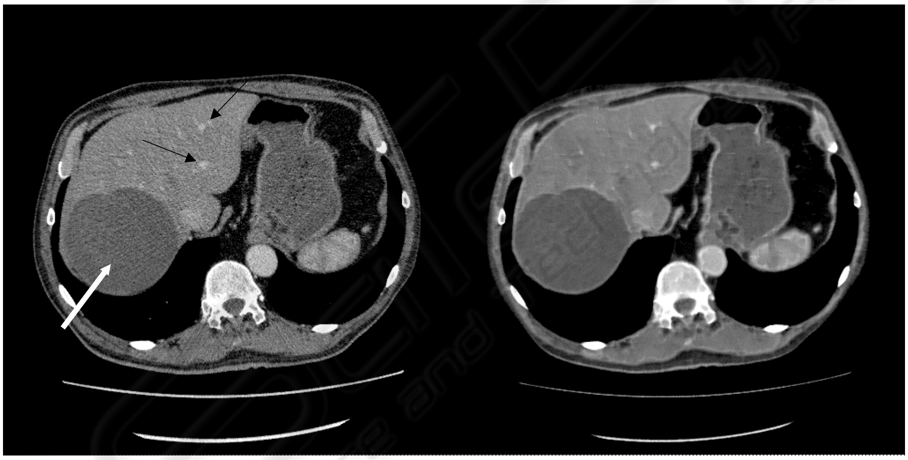

Fig. 2. Two-dimensional slices of tomography images before and after the processing at high level of the picture noising. The liver focal lesion (thick white arrow), the contrasted vessels (thin black arrows).

Some results of an image processing are presented in the paper. The Fig. 1 and Fig. 3 illustrate the results of the processing of tomography images with an average noise level. The most significant result is the vessels contrast enhancement. The utility of such a processing is rather better seen in 3D reconstruction. Due to significant noise reduction and contrast enhancement one can see greatly better vessels and kidneys macroanatomy details.

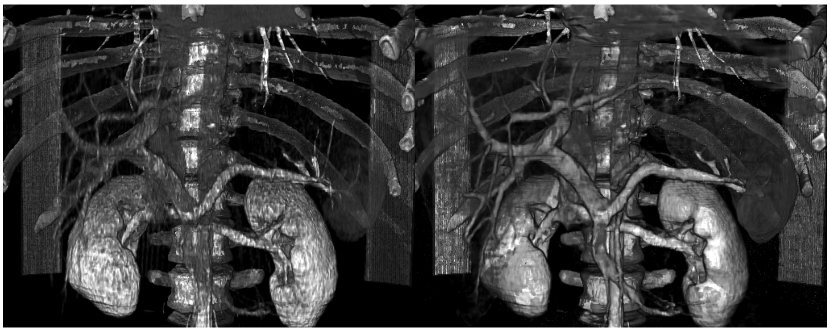

Fig. 3. The three-dimensional reconstruction of a tomography images before and after the processing at an average level of the picture noising. The contrasted liver vessels (portal vein, kidney veins).

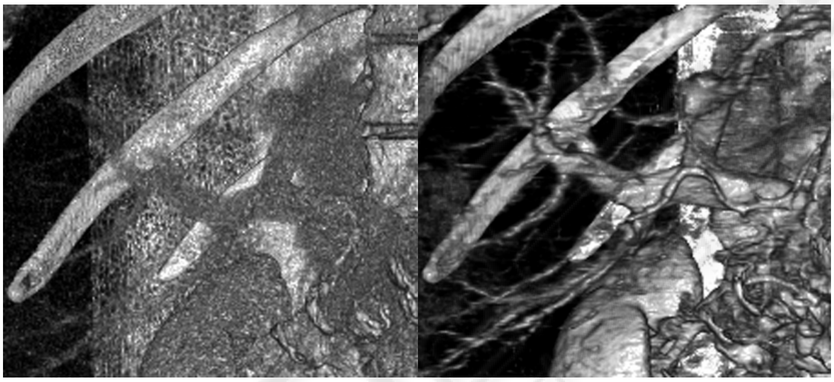

Fig. 4. The three-dimensional reconstruction of a tomography images before and after the processing at the strong level of the picture noising. The contrasted liver vessels.

Fig. 2 and Fig. 4 illustrate the processing of a strongly noised tomography images. In this case the analysis of even two-dimensional cross-sections may appear to be rather difficult, while in processed images the vessels are seen significantly better. The three-dimensional reconstruction of this tomography images has not yielded the desired results: the liver vessels either are not visualized at all or are presented by the dotted areas (what makes it impossible, in particular, to implement the evaluation of their volume). At reconstruction of the processed tomography images one can visualize vessels significantly more clearly; in this case they are visualized without breaks.

6 Conclusions

The elaborated technique allows an efficient suppressing the noise of computer tomography images and ensures a significantly improved three-dimensional reconstruction of the objects under examination.

The object "liver" and the case of 3D computed tomography images have been chosen just as an example. The application of 3D visualization is possible at any type of 2D tomography images (ultrasound, CT, MRT) and practically in any part of the human body (from head to foot).

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