Sygnatura: Pol J Radiol, 2007; 72(1): 65-70

Polish Journal of Radiology
www.PolRadiol.com
ORIGINAL ARTICLE

Otrzymano: 2006.03.23
Zaakceptowano: 2006.09.04

A new method of volume determination in three-dimensional ultrasonography

Grzegorz Tatoń[1], Eugeniusz Rokita[1], Zbisław Tabor[1], Jan Kulig[2], Marek Sierżęga[2], Stanisław Kłęk[2], Piotr Kołodziejczyk[2], Andrzej Urbanik[3]

[1] Department of Biophysics, Jagiellonian Univeristy-Collegium Medicum, Krakow, Poland
[2] I Department of General Surgery and Gastroenterological Surgery Clinic, Jagiellonian Univeristy-Collegium Medicum, Krakow, Poland
[3] Department of Radiology, Jagiellonian Univeristy-Collegium Medicum, Krakow, Poland

Author's address: Grzegorz Tatoń, Department of Biophysics, Jagiellonian Univeristy-Collegium Medicum, Krakow, Poland, Łazarza 16, 31-530 Kraków, e-mail: mmtaton@cyf-kr.edu.pl

The study sponsored by the Committee for Scientific Research within the research project 6 P05C 015 21

Summary

Background: The aim of this work was to develop a method of volume determination utilizing the grey scale histogram of three-dimensional (3D) ultrasonic (US) image. Volumes are calculated as the product of single voxel volume and the area of histogram peak representing investigated object. The proposed solution performance was compared with two other methods. First utilizes two-dimensional cross-section areas on subsequent image layers while the second allows the volume determination on the basis of one-dimensional measurements.

Material/Methods: The 3D US phantoms images were used to test the procedure. The usefulness of the method was also demonstrated on several clinical examples. The ultrasonic 3D images were collected, their histograms calculated and fitted with model curves allowing the volume calculations. The accuracy and precision was assessed and t-test was used for evaluating performance of all considered methods.

Results: The accuracy (understood as the difference between real and measured volume) achieved in the proposed solution (3.6%) was the highest comparing to alternative methods (5.2% and 8.4%). Also the p-value (two-tailed t-test) was better in the case of the presented method. The performance increase was due to the elimination of subjective delineation of measured regions of interest.

Conclusions: The presented procedure can be successfully used for volume assessment concerning its simplicity, accuracy and time consumption. It should be emphasized that the new method does not require image segmentation, unlike other methods in use. The procedure was tested with 3D US imaging but can be used successfully with any 3D imaging modality.

Key words: 3D imaging • 3D ultrasound • volumetry • image histogram analysis

PDF file: http://www.polradiol.com/fulltxt.php?ICID=468400

Background

The three-dimensional (3D) imaging was introduced into the medical practice in the last decade and its clinical usefulness is still growing [1–3]. Volume determination seems to be the most natural application of 3D imaging [1–5]. Most of the methods applied for volume determination employ the object delineation [6–8]. Separation of different anatomical and pathological structures on the ultrasonic (US) image (image segmentation) is challenging in the diagnostic imaging even in two dimensions (2D). There are no fully reliable automatic methods for image segmentation while manual segmentation is always subjective [1, 4, 6, 7, 9, 10].

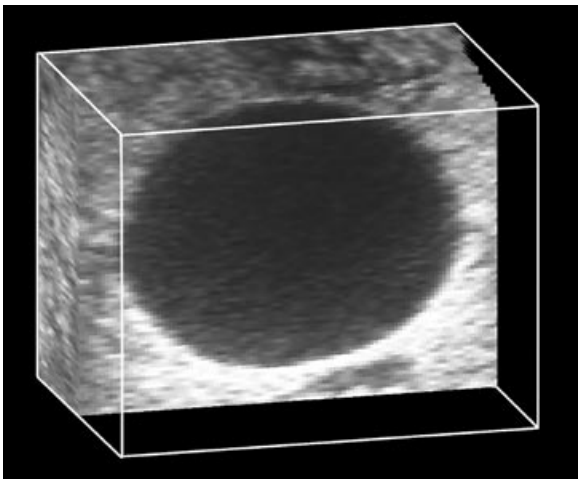

Figure 1. A sample of the three-dimensional image of the phantom analyzed in the presented study.

We present an alternative method for the evaluation of volume, which does not require object segmentation. The proposed method utilizes a 3D image histogram.

Materials and methods

Although the proposed method can be applied for 3D images acquired in any diagnostic modality (e.g. computed tomography or magnetic resonance imaging) in the presented study it was limited to the 3D US images. Nine phantoms in form of rubber balloons filled with water were prepared. They were embedded in kefir (curd) in order to simulate hypoechoic lesions within hepatic tissue. The Archimedes principle [4, 8, 11] was used to estimate the true volumes ranging from (11.7 ± 0.5) cm^3 to (91.8 ± 0.6) cm^3.

Several clinical data were involved to demonstrate the method performance in possible clinical applications. Informed consent was obtained from all subjects prior to performed investigations. The 3D images of 1 liver cyst, 1 pancreatic metastasis and 3 colorectal metastases were analyzed. In clinical examinations the measured volumes were compared to the results of volumetric measurements with computed tomography (CT) considered to be the best method to estimate the real volumes [5, 12]. CT examination was done with the use of Siemens Somatom Sensation 10 multislice spiral unit. The investigations taken for the analysis were characterised by the slice thickness of 2 mm reconstructed every 1.5 mm.

The US examinations were performed with Hitachi EUB-525 unit. The device was equipped with 3D imaging system with magnetic positioning [3, 4, 7, 8] delivered by Echotech 3D Imaging Systems (Hallbergmoos, Germany). Data pre-processed by the 3D imaging system software were exported as a series of TIFF files. The applied software provided calibrated 3D image resolution allowing

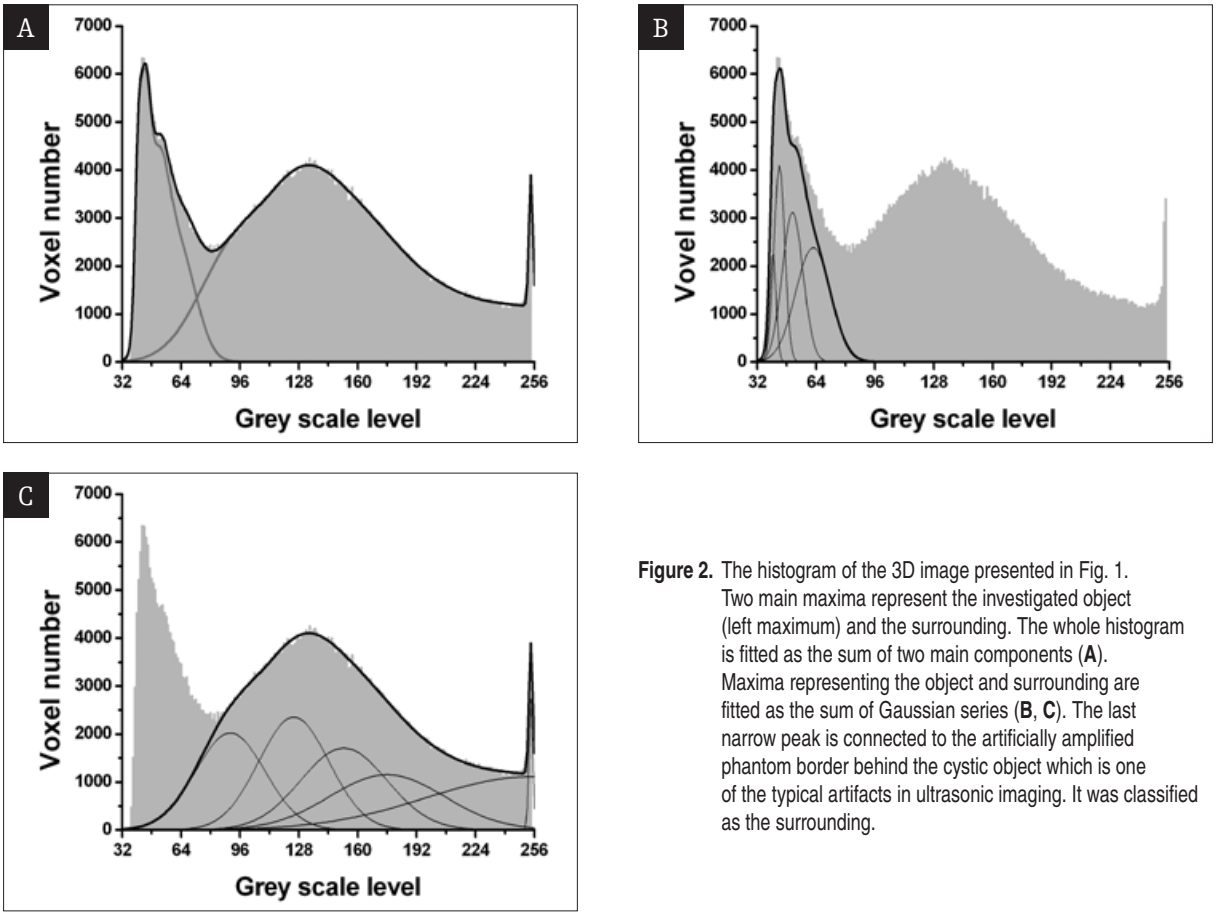

Figure 2. The histogram of the 3D image presented in Fig. 1. Two main maxima represent the investigated object (left maximum) and the surrounding. The whole histogram is fitted as the sum of two main components (**A**). Maxima representing the object and surrounding are fitted as the sum of Gaussian series (**B**, **C**). The last narrow peak is connected to the artificially amplified phantom border behind the cystic object which is one of the typical artifacts in ultrasonic imaging. It was classified as the surrounding.

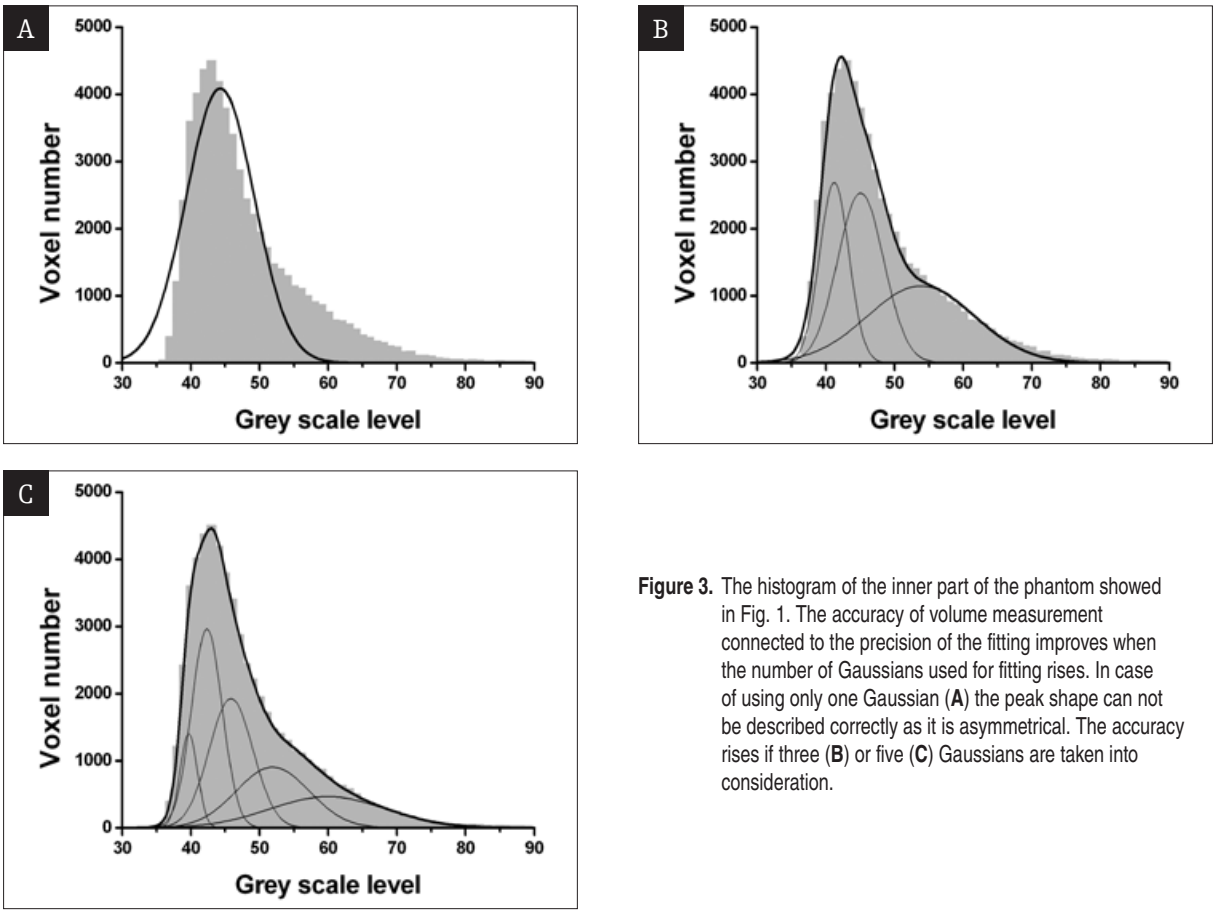

Figure 3. The histogram of the inner part of the phantom showed in Fig. 1. The accuracy of volume measurement connected to the precision of the fitting improves when the number of Gaussians used for fitting rises. In case of using only one Gaussian (**A**) the peak shape can not be described correctly as it is asymmetrical. The accuracy rises if three (**B**) or five (**C**) Gaussians are taken into consideration.

quantitative measurements. Data exported as a series of images were then imported and processed by the graphical workstation. DELL Precision Workstation 530MT equipped with two Intel Xeon 2.8GHz processors and 1GB of RAM was used. Special software created in our laboratories was used for data import. Additional procedures were added to the software for volumetric analysis described in this study.

The proposed method (HistMet) of volume calculation described below is based on the 3D image histogram which is a dependence of voxel number on voxel grey level. A block of 3D US image containing the interested objects (Fig. 1) was cut. The image was not filtered and it was cut as close to the object borders as possible in order to minimize the block volume.

Next, the histogram of the region containing measured object was calculated and analyzed. The method relies on the assumption that every distinguishable region in the image is reflected in the image histogram as a maximum. The peaks connected to different regions overlap in all cases because the investigated object and the surrounding contain voxels with the same grey level. In cases it is common to apply threshold segmentation but the proposed method is different. There is no need to delineate the investigated object, and it is not necessary to investigate where the overlapping voxels are. It is assumed that every object visible in the image gives certain continuous, peak-shaped distribution in the histogram. The proposed metod relies on the model curves fitting to histogram maximas representing all objects visible in the investigated region.

In all cases a single hypoechoic object embedded in the hyperechoic environment was observed. In the histogram it was presented as two separate peaks (Fig. 2): first – connected to the investigated object and second – coming from the surroundings. The opposite situation is also possible [5] but was not observed in the analysed data.

The area under the peak is equal to the number of voxels within the region represented by the maximum; therefore the area of the first maximum multiplied by the volume of single voxel gives the volume of the object. In order to calculate the area of first maximum the whole histogram was fitted to model curve being a sum of a few Gaussians. The area of peak was calculated as a sum of Gaussian areas contributing to the maximum (Fig. 2b). Those Gaussians were chosen on the basis of the inner object region histograms (Fig. 3) calculated separately.

The initial Gaussian parameters were first chosen manually and then the automatic fitting procedure was applied. The sum of square differences (SSD) between the fitted curve (the sum of all Gaussians) and the experimental data was minimized. SSD is a parameter widely used in the procedures aiming to the experimental data fitting [13]. The amplitude of SSD reflexes the fitting quality – the smaller the SSD value, the closer the model curves to the experimental data. SSD was normalized by the number of

Table 1. The result of measurements for phantoms (Ph1-Ph9) and clinical data (C1-C5) with the proposed method (HistMet) and two other volumetric methods (see text for more detailed description). All results are expressed in cm^3.

	Real volume	HistMet	SurfMet	2DMet
Ph1	11.7 ± 0.5	11.5 ± 0.6	12.2 ± 1.1	12.3 ± 1.4
Ph2	13.2 ± 0.5	13.3 ± 0.4	13.2 ± 1.2	14.1 ± 1.5
Ph3	18.3 ± 0.5	18.8 ± 0.6	19.9 ± 1.5	20.4 ± 1.9
Ph4	20.2 ± 0.5	19.7 ± 0.9	20.8 ± 1.5	22.2 ± 2.0
Ph5	23.2 ± 0.6	22.4 ± 0.7	23.4 ± 1.6	22.5 ± 2.0
Ph6	26.3 ± 0.6	25.7 ± 1.0	26.4 ± 1.7	27.2 ± 2.3
Ph7	46.7 ± 0.8	47.5 ± 4.4	46.6 ± 2.8	41.4 ± 3.0
Ph8	54.3 ± 0.9	55.4 ± 5.0	54.9 ± 2.4	51.0 ± 3.5
Ph9	91.8 ± 1.4	89.7 ± 3.3	95.0 ± 5.1	87.2 ± 5.1
C1	0.50 ± 0.09	0.51 ± 0.04	0.43 ± 0.07	0.47 ± 0.13
C2	0.62 ± 0.07	0.63 ± 0.04	0.62 ± 0.14	0.63 ± 0.15
C3	8.1 ± 0.6	8.2 ± 0.7	8.7 ± 1.3	6.6 ± 1.4
C4	10.8 ± 0.8	8.8 ± 0.3	12.4 ± 1.8	11.1 ± 1.6
C5	14.9 ± 0.6	16.0 ± 0.6	17.0 ± 1.6	18.8 ± 1.8

histogram points and by the number of image points [13]. The fitting procedure was performed iteratively until SSD amplitude was stable, what took several minutes for the computer involved in the calculations.

After the fitting procedure was complete, parameter errors were estimated. We applied the approach typical of minimizing the problems [13]. The value of every fitted parameter was changed step by step and the SSD changes were analyzed. The value which caused 1% change of SSD was considered as a parameter error [13]. Volume uncertainty was estimated by error propagation associated with parameters of the model curves as the volume was calculated from the fitted model curves.

The HistMet method was compared to two other volumetric methods based on the object border delineation. The first method (SurfMet), used very often in practice, calculates the object volume from the object cross-sectional areas measured on subsequent layers [4, 6, 7, 11]. The areas, multiplied by the layer thickness, were added to give the object volume. The cross-sectional areas were defined manually. In order to estimate the method uncertainty it was assumed that the cross-section border defined manually can be shifted by one pixel. The approximate error of the cross-sectional area is equal to the object circumference multiplied by the pixel size. The error of the estimated volume is calculated using the errors of all cross-sectional areas calculated on subsequent layers.

Table 2. Results of t-test applied for performance assessment of all discussed methods. The hypothesis that the measurement results and the real volumes for phantom studies, or volumes measured with computed tomography for clinical data, are significantly different was tested. The obtained p-values suggest that the hypothesis is false. For details – see the text.

	p-value	
	Phantom studies	Clinical data
HistMet	0.988	0.970
SurfMet	0.953	0.850
2DMet	0.945	0.908

The second method (2DMet) simulates the process of volume assessment on the basis of clinical 2D US. There is a particular object shape assumed and its dimensions are measured on the 2D US images. The volume is calculated with the use of mathematical formulas for the specific shapes. For our measurements it was assumed that the objects shapes can be estimated by an ellipsoid [11, 14, 15]. The object axes in three orthogonal directions were measured and volumes were calculated. Uncertainty of the method was calculated on the basis of errors in measured axes lengths, assumed to be equal to double pixel size.

The results obtained from all three methods were compared with the real values (in the phantom studies) or with CT results (clinical data) using two tailed t-test for correlated samples. The precision and accuracy of all measurements as well as the mean precision and accuracy for all three methods were calculated.

Results and discussion

The detailed results of phantom and clinical data measurements are presented in Table 1. The results of t-test applied

Table 3. The mean precision (%) achieved in the discussed methods. For abbreviations – see the text.

	Phantom studies	Clinical data	All
HistMet	5.0	6.0	5.3
SurfMet	6.9	15.5	10.0
2DMet	8.6	19.3	12.5

Table 4. The mean accuracy (%) of the discussed methods. For abbreviations – see the text.

	Phantom studies	Clinical data	All
HistMet	2.2	6.1	3.6
SurfMet	2.4	10.1	5.2
2DMet	6.9	11.0	8.4

for all three methods are presented in Table 2 while the calculated mean precisions and mean accuracies are given in Tables 3 and 4.

The t-test results and accuracy calculations confirm that the presented method is more efficient than the currently used methods of volume determination. The advantage is especially visible for clinical data as the delineation of observed object is difficult [4, 5]. The method based on the 3D image histogram analysis is characterised by higher accuracy. The use of automation reduces the subjective factor playing the main role in manual object segmentation [5, 9, 10] which is indirectly involved in both SurfMet and 2DMet methods. On the other hand, automatic segmentation is problematic due to the difficulties with low reliability of automatic segmentation procedures in the case of diagnostic images [5].

The worst results were obtained from the volume measurements on the basis of 2D US (the lowest accuracy). The 2DMet should be considered as a method of volume estimation rather than determination. Accepting difficulties of linear dimensions measurements connected to the indirect segmentation the theoretical shape assumption for volume calculation precludes the accurate results [14].

As expected, the performance of all methods is worst when clinical data are taken into consideration. It is more complicated to perform measurements in real clinical conditions than in the situations when conditions are well defined and controlled, e.g. in case of phantom investigations. This obvious conclusion is in agreement with the reported data [5–8].

The method using the image histogram for volume determination eliminates the subjective manual or not reliable automatic segmentation procedures. It is especially useful in the 3D US imaging where the segmentation is usually difficult [4, 5, 9, 10] or the object shape is complicated. The problem of segmentation is replaced by other numerical problem – minimization [13]. The model curve needs to be fitted to the histogram. The sum of Gaussian curves was applied as it is the most popular and accessible in programs used for data analysis. On the other hand, single peak in the US image histogram is asymmetrical and can not be fitted with single Gaussian curve (Fig. 3). The number of model curves and their classification is disputable. As shown in Fig. 3, the number of Gaussians fitted to the peak determines the quality of fitting. The accuracy significantly improves with the number of used curves, what was confirmed quantitatively (Fig. 4). However, the number of Gaussians used in fitting procedure increases the amount of time necessary for calculations. The time required for 100000 iterations in the minimization procedure was about 28 s when 5 Gaussians were used and about 52 s in the case of 10 Gaussians. The dependence between time-consumption and number of Gaussians is linear. The number of the applied Gaussians is a compromise between accuracy and efficiency. Ten curves were used in all cases as this number gave reasonable accuracy and time consumption. It was the smallest number of Gaussians enabling the match with SSD<1 for all investigated data.

Other important problem is the identification of model curves contributing to particular peak in the histogram (Fig. 2). It is simple as long as the maxima representing object and surroundings are well separated but that is not always the case, especially in clinical data. The problem was solved by fitting the histogram of the inner part of measured object. Such approach allowed the identification of possible Gaussian positions when the whole histogram was fitted. The number of Gaussians contributing to the peak representing the object in the image histogram can differ according to different cases depending on the results of fitting procedure. Usually four or five Gaussians were used to approximate the peak shape.

The use of minimization instead of segmentation cannot eliminate the subjectivity of measurements because some parameters used in minimization procedure have to be set by the program operator (e.g. the level on which the SSD parameter is stable, the number of iterations or the SSD variation level in the procedure of uncertainty assessment). The problem was solved by setting the variable parameters on the same level for all measurements.

The accuracy of the method was evaluated on the basis of subjective parameter controlling the possible SSD change

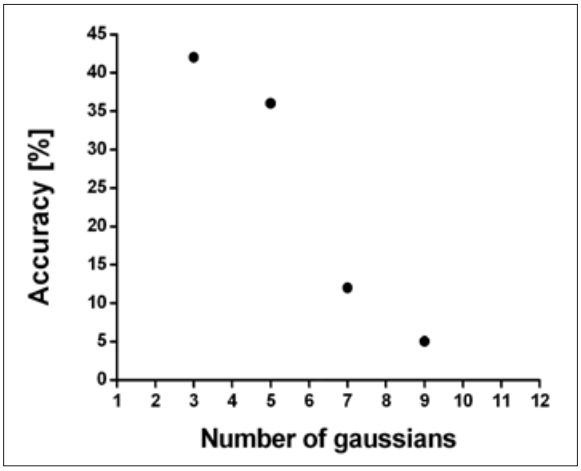

Figure 4. The dependence between volume measurement accuracy and number of Gaussians used for the image histogram fit. Calculations performed for phantom showed in Fig. 1.

caused by the Gaussians parameter errors as described earlier. The precision is calculated subsequently by the error propagation. The SSD change level could be established on the value other than 1%. There is no objective method to calculate the error minimization problems [13]. The order of magnitudes of calculated accuracies and precisions in phantom studies are the same, what suggests that the choice was done properly.

The precision and accuracy of HistMet method depends on the size of cut 3D image block containing the investigated object. When the block size is bigger, more surrounding tissue is represented by larger maximum which strongly overlaps the peak representing the object. This effect was not investigated quantitatively – instead it was only minimized by cutting as small image block as possible. Such approach limits as well the risk of including objects in the 3D field of view other than the investigated object and surroundings. In such situation more peaks would have been observed and the analysis would have been more complicated, yet possible.

If many objects with different echogenities were visible in the region of interest, volumes of all could be measured simultaneously, which would undoubtedly stand for an advantage. In such case more model curves should be applied to fit the image histogram. Another possibility is to use other model curves for fitting particular peaks instead of the Gaussian series. On the other hand, the presented method can fail in case when there are two or more objects characterised by the same echogenity but separated. It would be possible to measure the total volume of all such objects but the volume of one particular could not be measured.

Most of the artefacts typical for US images will be present in the image histogram as additional peaks or will influence the measurements results in some other way. An example is shown in Fig 2. The last narrow peak is connected to the object border amplification (Fig. 1) and not proper US machine settings. The quantitative assessment of different artefacts influence on the results requires further, more detailed investigations.

References:

1. Campani R, Bottinelli O, Calliada F, et al.: The latest in ultrasound: three-dimensional imaging. Part II. Eur J Radiol, 1998; 27 supl 2: S183–S187.
2. Downey DB, Fenster A, Williams JC: Clinical utility of three-dimensional US. RadioGraphics, 2000; 20: 559–571.
3. Nelson TR, Downey DB, Pretorius DH, et al. Three-dimensional ultrasound. Philadelphia: Lippincott Williams & Wilkins; 1999.
4. Matre K, Stokke EM, Martens D, et al.: In vitro volume estimation of kidneys using three-dimensional ultrasonography and position sensor. Eur J Ultrasound, 1999; 10: 65–73.
5. Park SH, Choi BI, Han JK et al.: Volumetric tumor measurement using three-dimensional ultrasound: in vitro phantom study on measurement accuracy under various scanning conditions. Ultrasound Med Biol, 2004; 30: 27–34.
6. Farrell T, Leslie JR, Chien PFW, et al.: The reliability and validity of three dimensional ultrasound volumetric measurements using in vitro balloon and in vivo uterine model. British J Obstretics Ginecology, 2001; 108: 573–582.
7. Hughes SW, D'Arcy TJD, Maxwell DJ et al.: Volume estimation from multiplanar 2D ultrasound images using a remote electromagnetic position and orientation sensor. Ultrasound Med Biol, 1996; 22: 561–572.
8. Strommen K, Stormark TA, Iversen BM, et al.: Volume estimation of small phantoms and rat kidneys using three-dimensional ultrasonography and a position sensor. Ultrasound Med Biol, 2004; 30: 1109–1117.
9. Polakow J, Janica J, Serwatka W, et al.: Value of three-dimensional sonography in biopsy of focal liver lesions. J Hepatobiliary Pancreat Surg, 2003; 10: 87–89.
10. Thune N, Gilja OH, Hausken T, et al.: A practical method for estimating enclosed volumes using 3D ultrasound. Eur J Ultrasound, 1996; 3: 83–92.
11. Partik BL, Stadler A, Schamp S et al.: 3D versus 2D ultrasound: accuracy of volume measurements in human cadaver kidneys. Invest Radiol, 2002; 37: 489–495.
12. Klek S, Kulig J, Popiela T et al. The value of modern ultrasonographic techniques and computed tomography in detecting and staging of pancreatic carcinoma. Acta Chir Belg, 2004; 104: 659–667.
13. Press WH, Teukolsky SA, Vetterling WT, et al.: Numerical recipes in C. The art of scientific computing. Cambridge: Cambridge Univerity Press; 1992.
14. Bakker J, Olree M, Kaatee R et al.: Renal volume measurements: accuracy and repeatability of US compared with that of MR imaging. Radiology, 1999; 211: 623–628.
15. Wolf GK, Lang H, Prokop M, et al.: Volume measurements of localized hepatic lesions using three-dimensional sonography in comparison with three-dimensional computed tomography. Eur J Med Res, 1998; 23: 157–164.