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Phantom studies as useful method for optimization of CT procedures with single-slice scanners

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Background:

Optimization of X-ray diagnostic procedures requires a selection of exposure parameters adequately for the patient anatomical structure and clinical needs, meanwhile use of the ready software protocols is a daily routine.

The paper presents the methods and the results based on phantom studies aimed to optimize the exposure parameters for the examined object in examinations performed using single-slice CT scanners.

Material/Methods:

The studies were concerned with the significance of particular exposure parameters for image quality and patient doses in routine head procedures. Image quality was evaluated using a Catphan 424 phantom. Dosimetric measures of exposure were evaluated by computed tomography dose index (CTDI), and dose-length-product (DLP) was measured using a Nomex dosimeter.

Results/Conclusions:

Two single-slice CT scanners were investigated in details: a fourth generation one (Picker PQ-2000) and a third generation one (Siemens Somatom Balance). The dependence of image quality factors on high voltage, anode current, scan time, slice thickness were described and CTDI was evaluated.

For both the CT scanners tested, the combinations of exposure parameters were found for which low dose value was accompanied by good image quality.

Moreover, the high-contrast resolution was not practically changed with dose reduction, in opposite to low-contrast resolution being related to exposure parameters and the type of CT scanner. Dose reduction possible for the fourth generation scanner is deeper.

Key words:

computed tomography • phantom studies • image quality

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Background

The importance of X-ray computed tomography (CT) as an imaging technique has increased immensely during the 25 years which elapsed since the development of the first tomographs. X-Ray tomography is now considered to be the method of the highest diagnostic value among the available radiodiagnostic techniques.

Increasing popularity of CT has led to a rise in population exposure to X radiation, which has been emphasized twice by the European Commission in the published guidelines

discussing the fundamentals of correct technique in most common CT examinations and the method of dosimetric assessment of exposure [1, 2], the International Committee for Radiological Protection in Publication [3], as well as by the International Atomic Energy Agency, which initiated a research program aimed at the attempt to work out the methods of dose restriction. The problem of increasing patient exposure to radiation during CT procedures is present also in Poland: during the period from 1995 to 2004, the annual number of CT examinations increased five-fold, and, according to recent statistics, exceeds 1100 thousand at present [4, 5].

Despite installation of many multi-row scanners during the recent years, numerous centers still use older types of single-row tomographs. In view of increasing degree of wear and tear of this equipment, the search for optimization of the procedures, by determining the possibility of obtaining diagnostically satisfactory visualization of the object with reasonable limitation of the radiation dose received by the examined patient, is particularly important. Such optimization is required now according to legal regulations [6], and in case of single-row scanners it is entirely the user's responsibility, because no software solutions in this field have been provided by the manufacturers. Optimization of diagnostic procedures requires individual solutions for each scanner type, based on testing its visualization potential with respect to standard examination objects.

For this purpose, phantom studies have been designed, the methods and results of which used and obtained for third and fourth generation single-row scanners are presented in this paper.

Optimization of CT procedures involves obtaining images of diagnostically satisfactory quality maintaining the dose of radiation at a reasonably low level. The image quality is determined by three basic indexes: spatial resolution, contrast level and noise level.

Spatial resolution is the potential to differentiate – as separate ones – two linear structures characterized by a high attenuation coefficient (so called high-contrast objects). This resolution is deteriorated when the size of tube focus increases (with progressive wear of the x-ray tube), or in the presence of artifacts, e.g. scatter of the radiation on elements absorbing high X-ray doses (e.g. implants). The effect of spatial resolution deterioration may be due also to too high (in relation to the properties of the examined object) anode current in the scanner tube.

Contrast in CT images – or contrast resolution – can be interpreted in the simplest way as visualization of the particular object on the background of water (nominal 0 HU). Thus, contrast level is determined by the values of attenuation coefficients, which, in turn, are dependent on energy. Consequently, the contrast level is dependent on the value of voltage between the X-ray tube electrodes.

Noise level is another important determinant of image quality. Its main component is so-called pixel noise [1], which is a result of statistical difference of CT numbers in the elements of the image. This difference increases with an increase of scanning beam intensity, which, obviously, is dependent both on the transmitted charge and on the voltage between the X-ray tube electrodes. The function

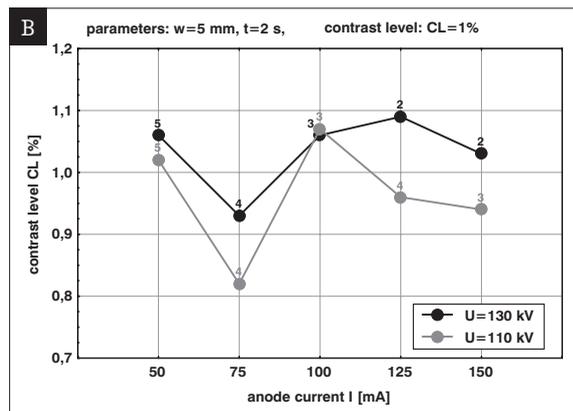
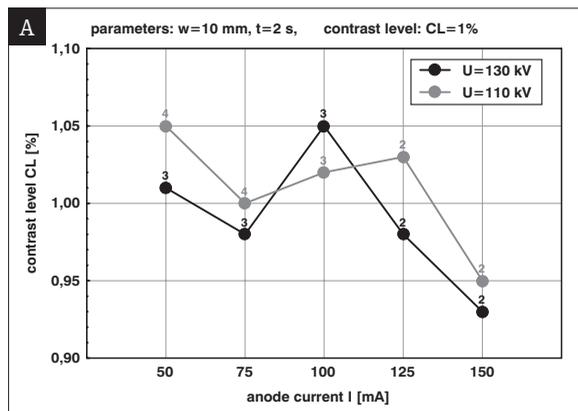


Figure 1. Third generation scanner (Siemens Somatom Balance). Relationship between contrast level and anode current for objects of nominal contrast level 1%, at voltage of 130 kV and 110 kV: A. 10 mm slice thickness, B. 5 mm slice thickness.

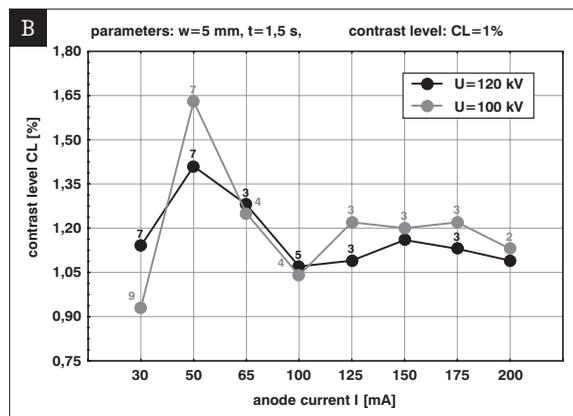
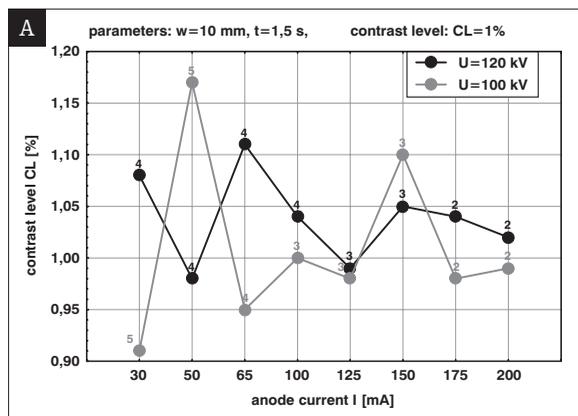


Figure 2. Fourth generation scanner (Picker PQ2000). Relationship between contrast level and anode current for objects of nominal contrast level 1%, at voltage of 120 kV and 100 kV: A. 10 mm slice thickness, B. 5 mm slice thickness.

dependence form is determined by the emission output of the scanner, but also by the quality of its detectors and the installed software.

However, it can be demonstrated on the basis of theoretical calculations that for most soft tissue objects (with contrast level higher than 1%), despite increased noise due to a reduction of voltage, their visualization is not deteriorated, and the radiation dose decreases [7, 8]. Although it means optimization of the procedure, modification of the protocols in clinical examinations performed using a particular scanner must be preceded by phantom studies, which allow to verify the expectations with respect to that scanner. The results of such studies are presented in this paper.

Materials and methods

In the studies conducted within the framework of this research project, the indexes of CT image quality were determined using a Catphan 424 phantom (The Phantom Laboratory Incorporated, Salem, NY), designed for the control of quality of single-row scanner with axial mode of function.

The low-contrast section of this phantom consists of four eight-element groups with nominal contrast level differences of 1%, 0.5%, 0.3% and 0.1%, and each group consists of elements from 2 to 15mm in diameter.

The high-contrast section of this phantom contains a „ridge“ consisting of groups of lines, corresponding to the resolution of 5 to 20 line pairs per centimeter. A detailed description of the phantom is presented in papers [9, 10].

Dosimetric assessment of exposure made use of indexes consistent with the European Commission guidelines [1], according to which patient exposure to radiation during CT procedures is determined by mean dose per scan (CTDI_{vol}) index, expressed by the following formula:

$$CTDI_{vol} = \frac{DLP}{w \cdot p}$$

where:

$$\overline{DLP} = \frac{1}{n} \sum_{i=1}^n \left(\frac{1}{3} \cdot DLP_i^c + \frac{2}{3} \cdot DLP_i^p \right)$$

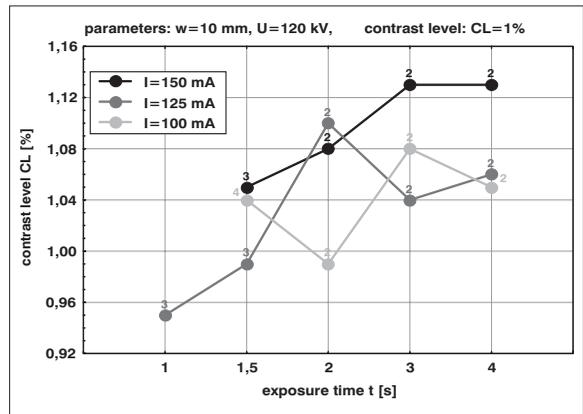
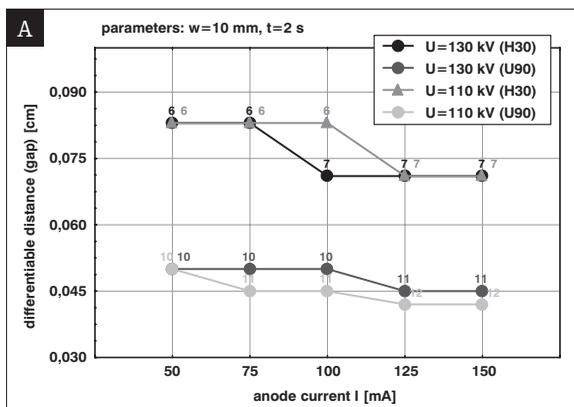


Figure 3. Relationship between contrast level and scan time for objects with nominal contrast level 1%, at 10 mm slice thickness, 120 kV and different anode current values.

- DLP* – product of the dose and length of the scanned segment (dose length product),
- DLP_i^c – DLP value in the phantom center in the „i“-th scanned layer,
- DLP_i^p – DLP value on the phantom periphery in the „i“-th scanned layer,
- w* – scanned layer thickness,
- p* – pitch,
- n* – number of layers scanned.

The starting point for the analysis was the routine head CT scanning protocol. The parameters were modified within the range available with the scanner software and accepted by the scanner control system.

Results

Detailed studies were carried out for two single-row computed tomographs: a third generation Siemens Somatom Balance (SSB) scanner and a fourth generation Picker PQ-2000 (PQ) one.

The studies included the assessment of effect of the individual exposure parameters on the basic image quality indexes and the mean dose per scan (CTDI_{vol}) value. The complete

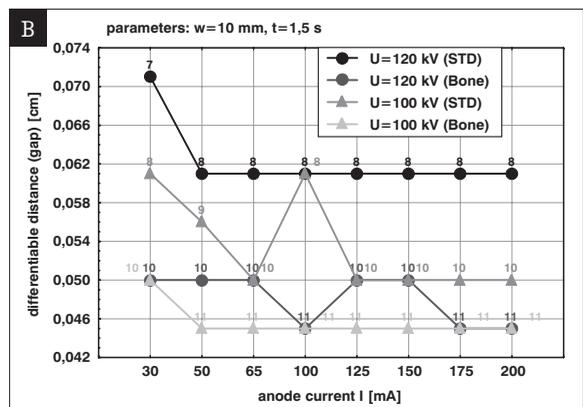


Figure 4. Detectable distance between high-contrast objects as function of anode current at 10 mm slice thickness: **A.** III generation scanner – at 130 kV and 110 kV for standard (H30) and bone (U90) filter, **B.** IV generation scanner – at 120 kV and 100 kV for standard (STD) and bone (BONE) filter.

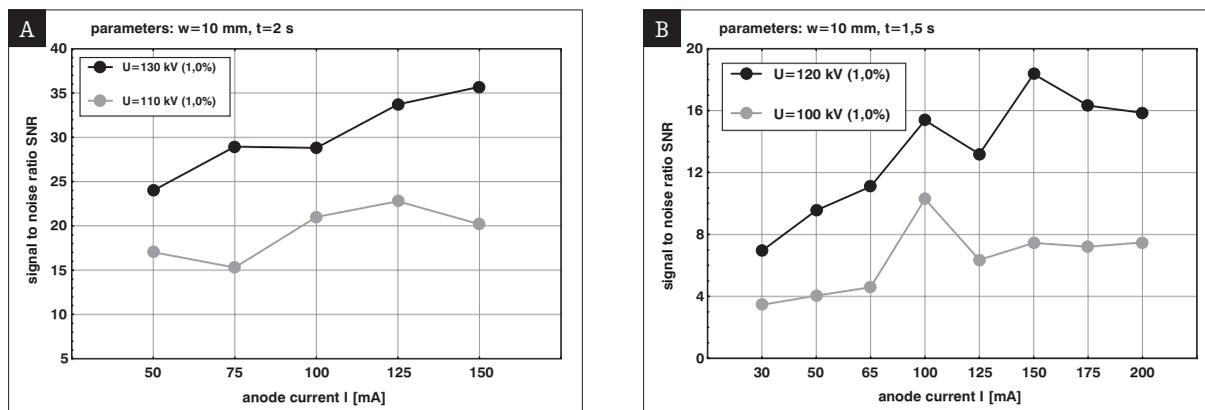


Figure 5. SNR as function of anode current for objects with nominal contrast level 1% at 10 mm slice thickness: **A.** III generation scanner – at 130 kV and 110 kV, **B.** IV generation scanner – at 120 kV and 100 kV.

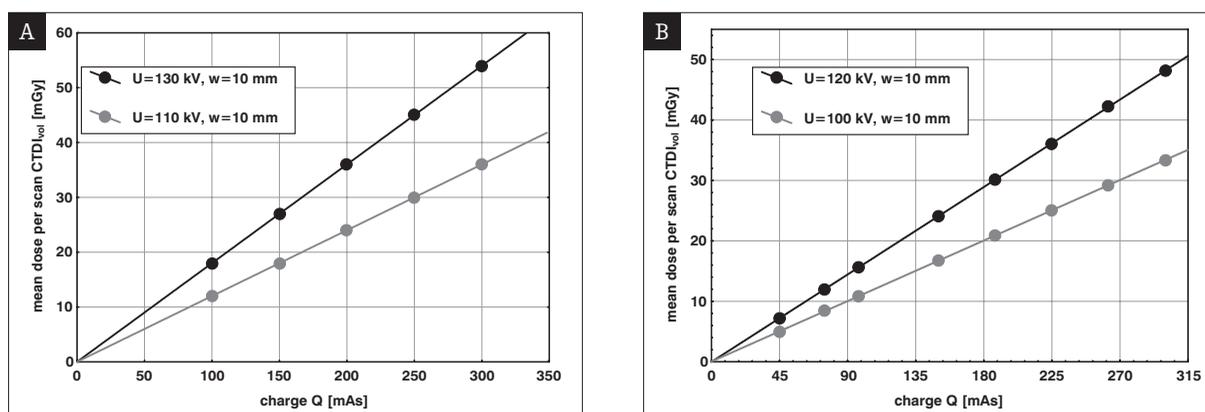


Figure 6. Mean dose per scan ($CTDI_{vol}$) as function of tube output (Q) at 10 mm slice thickness: **A.** III generation scanner – at 130 kV and 110 kV, **B.** IV generation scanner – at 120 kV and 100 kV.

results have been published in another paper [10]. Because of obvious limitations of this publication, the results are presented for objects with the nominal contrast level of 1%, i.e. corresponding to most soft tissue objects. CT image contrast level of 1% guarantees clear visualization of extravascular blood collections on the background of neural brain tissue.

Dependence of the estimated contrast level on anode current intensity for 2 voltage values available with the possessed SBB scanner software is presented in fig. 1. An analogical dependence for the PQ scanner is presented in fig. 2. As the PQ scanner software made it possible to adjust the single scan time (independently of current intensity), the results of the respective measurements are presented in fig. 3.

Dependence of differentiable distance of high-contrast objects (i.e. spatial resolution) on anode current intensity is presented for both scanners in fig. 4.

Signal-to-noise ratio (SNR) is another frequently used index of image quality. The SNR values for the analyzed phantom images in relation to anode current intensity are presented in fig. 5.

Mean dose per scan ($CTDI_{vol}$) values in relation to anode current and single scan time product are presented for both scanners in fig. 6.

Discussion

The results of thorough analysis of correlations between image quality indexes and exposure parameters, collected as a result of tests carried out within the framework of this study make it possible to compare the reactions of third and fourth generation single-slice scanners. The observations concerning the essential indexes of image quality have been presented in tables 1, 2 and 3. Table 1 presents the range of reduction of exposure parameters that is possible to obtain and its effect on low-contrast resolution of both scanners (Table 1). Table 2 illustrates the effect of anode current intensity variations on the range of signal-to-noise ratio (SNR) (Table 2). As it is known, the value of computed tomography dose index (CTDI) changes in proportion to anode current intensity. Thus, by relating the information concerning the proposed range of current intensity changes with the corresponding changes in signal-to-noise ratio (SNR) makes it possible to assess how much the dose can be reduced to maintain satisfactory image quality. Table 3 shows the data concerning the relation between SNR reduction ratio (G_{SNR}) corresponding to the reduction of I [mA] at which the diameter of the smallest detectable low-contrast object does not exceed $2 \times d_{min}$ visible when using the parameters specified in the manufacturer's protocol (Table 3).

Table 1. The range of reduction of exposure parameters and low-contrast resolution.

Contrast level CL [%]	Slice thickness [mm]	Picker PQ 2000 (Scanning time t= 1.5 s)			Siemens Somatom Balance (Scanning time t= 2 s)		
		U [kV]	d _{min} * [mm]	I [mA] reduction range until doubling of d _{min}	U [kV]	d _{min} * [mm]	I [mA] reduction range until doubling of d _{min}
1.0	10	120	2	200→30	130	2	150→50
		100	2	200→65	110	2	150→50
1.0	5	120	2	200→65	130	2	150→75
		100	2	200→125	110	3	150→50

*diameter of the smallest detectable low-contrast object in the images of Catphan 424 phantom obtained with current intensity specified in the manufacturer’s protocol.

Table 2. The range of SNR corresponding to the whole range of the available values for 1% contrast objects.

Slice thickness [mm]	Picker PQ 2000		Siemens Somatom Balance	
	Scanning time t=1.5 s; Current intensity range I= (30-200) mA		Scanning time t=2 s; Current intensity range I= (50-150) mA	
	U[kV]	Range SNR	U[kV]	Range SNR
10	120	7→17	130	20→35
10	100	3→7	110	13→25
5	120	6→14	130	17→25
5	100	3→8	110	10→15

Table 3. Lowering signal to noise (G_{SNR}) related to reduction of radiation (R_{mAs}) (to the level at which low-contrast resolution is twice worse (2 x d_{min})).

Picker PQ 2000			Siemens Somatom Balance		
U [kV] Slice thickness [mm]	R _{mAs}	G _{SNR}	U [kV] Slice thickness [mm]	R _{mAs}	G _{SNR}
120 kV; 10 mm	6.7	2.4	130 kV; 10 mm	3.0	1.7
100 kV; 10 mm	6.7	2.3	110 kV; 10 mm	3.0	2.0
120 kV; 5 mm	3.0	2.0	130 kV; 5 mm	2.7	1.4
100 kV; 5 mm	1.6	1.5	110 kV; 5 mm	4.0	3.0

Signal to noise reduction ratio (G_{SNR}) is defined as the quotient of values corresponding to the lowest and the highest values of the current intensity range within which d_{min} does not increase above the doubled baseline value.

Correspondingly – the change in intensity of the original beam was expressed as a quotient of the initial and the final charge values (I x t), corresponding to the above criteria of low-contrast resolution (2 x d_{min}), i.e. R_{mAs}.

(Such a method of presentation allows direct comparison of third and fourth generation scanners, which have different scanning times set in the standard head examination protocols).

The values presented in table 3 document the fact that PQ – as a generation IV scanner demonstrates, for 10 mm slice thickness, considerable flexibility of the primary scanning beam intensity; nearly 7-fold dose reduction results only in slightly over 2-fold deterioration of SNR, irrespectively of the voltage. For 5 mm slice thickness and 100 kV voltage this flexibility is lost: SNR decreases to the same extent as the dose is reduced.

Third generation scanners, due to their structure, have no such „imaging power reserves”, although the dose reduction ratio also exceeds that of SNR deterioration. Unlike the PQ scanners, SSB demonstrates better flexibility to modification of scanning parameters after voltage reduction to 110 kV.

Additionally, the dependence between low-contrast resolution and one scan duration was analyzed for PQ (Figure 3). It was found that for slice thickness w=10 mm high contrast level is guaranteed by Q=200 mAs, with d_{min} value lower when single scan duration is longer and current intensity higher than for the opposite combination of these elements; increase of anode current intensity (I) and scan time (t) did not cause significant improvement.

For voltage value U=100 kV, in order to obtain the same result as presented above, a charge value Q=250 mAs, obtained as a product of anode current I=125 mA and scan time t=2 s, is needed.

The presented result provides evident confirmation of the plausibility of optimization of CT procedures: *an appropriate*

Table 4. The best results of the high-contrast resolution for image obtained with two different scanners (slice width $w=10$ mm).

Scanner	U[kV]	Filter function	Anode current intensity range [mA]	Differentiable distance between lines [mm]
Picker PQ 2000	120	STD	50-200	0.62
		BONE	30-200	0.45-0.50
	100	STD	65-200	0.50
		BONE	50-200	0.45
Siemens Somatom Balance	130	H30	100-150	0.71
		U90	125-150	0.45
	110	H30	125-150	0.71
		U90	125-150	0.41

– and not the highest – number and X energy quantum allows to obtain the optimal image quality in CT procedures.

The considerations presented so far concern low-contrast resolution, as a problem more difficult in diagnostic practice. Nevertheless, the results obtained in the study include also data concerning high-contrast resolution, and their comparison between both scanners (generation III and IV) has been presented in table 4 (Table 4).

The data in table 4 indicate that high-contrast resolution is only slightly affected by modification of exposure parameters. The resolution of images obtained using the standard filter function is better for generation IV scanners but this difference disappears after image transformation with the function used for assessment of bone structures.

The conclusions based on the results presented above correspond with the studies published by other authors. However, it should be emphasized that *no* papers presenting the results of complex assessment of the effect of various exposure parameters on the particular image quality indexes in CT procedures have been found in the available literature. A few exceptions include position [11], the authors of which compared the relations between CTDI and low-contrast resolution for various scanners. The comparison concerned helical scanning mode with the parameters used for scanning of the liver (120 kV, 240-250 mA, 1s, slice thickness 5 mm). Low-contrast resolution was assessed on the basis of images of a 3D phantom designed at the University of Erlangen [12]. Although the results described in the cited paper [11] also concern two PQ 5000 scanners, their comparison with the results obtained in this study does not seem to be reasonable due to significant differences in methodology:

1. a different phantom used for assessment of resolution,
2. the region of interest (ROI) used in study [11] is nearly 7-fold largest, which results in artificial reduction of the recorded noise level,
3. in study [11], the phantom images were analyzed by an indirect method – after recording in a film form, and not on the computer screen, as in routine analysis of diagnostic images,
4. paper [11] does not mention whether all the images were analyzed using the filter function comparable for the particular scanners.

Nevertheless, it should be emphasized that the authors of [11] also signal the differences between generation III and IV scanners, although these differences are not analyzed in detail. A lot of information concerning the factors affecting the quality of CT images can also be found in the paper by Huda [13]. The opinions presented there can be summarized as follows:

1. the prerequisite for good image quality is the selection of exposure parameters (charge and voltage) appropriate for the diagnostic needs of the patient, taking into consideration the characteristics of the scanner used;
2. the most important value characterizing image quality is signal-to-noise ratio (SNR); changes of exposure parameters introduced during examination of a particular patients with a specific scanner type are acceptable if they do not cause a significant deterioration of SNR;
3. reduction of voltage is favorable especially when structures characterized by a high attenuation coefficient are imaged.

Huda [13] emphasizes also that the use of 80 kV voltage instead of 120 kV often used in pediatric head scanning procedures allows a significant dose reduction with preservation of good image quality.

The problem of optimization of CT scanning procedures has been often discussed in the literature during the recent years, but mainly with respect to multi-row scanners. Generally, it involves adjustment of X radiation scanning beam intensity to the absorption properties of the scanned element, which is effected automatically by the scanner program without the participation of the operator, in the real time of the scan. Such a procedure is realized in various ways by different manufactures of CT scanners [14, 15, 16].

Such a solution of the dose reduction problem is theoretically possible also for single-row scanners: such attempts were made in the late 1990's [17]; however, the solutions did not come into common use because they would significantly decelerate the process of data acquisition and would require a change of the software package, which would be associated with high costs. Single-row scanners are still being commonly used, but the problem of optimization of scanning procedures must be solved by the users on their own, on the basis of fragmentary information available in the literature.

Conclusions

The results of tests and measurements carried out within the framework of this study allow to present the following conclusions:

1. For single-slice CT scanners analyzed in this study, it is possible to reduce considerably the current- and voltage-related exposure parameters maintaining satisfactory image quality. The level of reduction decreases within decreasing thickness of the scanned layer.
2. A range of exposure parameter reductions is possible in head CT procedures because of considerable differentiation of anatomic structures with respect to their X-radiation attenuation properties.
3. Reduction of current- and voltage-related exposure parameters exerts different effects on the essential image quality indexes, i.e. low- and high-contrast resolution:

- high-contrast resolution is only minimally changed as a result of modification of exposure parameters
- low-contrast resolution changes are specific for particular scanner types and markedly dependent on current- and voltage-related exposure parameters and slice thickness. Low-contrast resolution can be improved more easily by increasing the single scan time than by increasing the intensity of anode current, maintaining the product of these parameters ($I \times t$ [mAs]), i.e. the value of the charge flowing through the scanner lamp, constant.

For a specific scanner type the charge value (Ixt [mAs]) at which maximum image quality is reached can be determined; further increases of this parameter are associated with no benefits, but only with unnecessary dose increases.

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