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Influence of exposure parameters on patient dose and image noise in computed tomography

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This work presents the results of investigation of influence of main exposure parameters on the patient dose and image noise on five clinical computed tomography units. Patient dose was determined by means of the quantity $CTDI_{air}$ free on air measured with pencil shaped ionization chamber. The image noise was estimated as a standard deviation of CT numbers in a 500 mm^2 central region of interest in a water-equivalent phantom positioned in the centre of rotation. The alteration of tube voltage and tube current show the potential in patient dose reduction up to 40% but may deteriorate image quality and has to be carefully applied in order to optimize the clinical CT protocols taking into account the specific clinical task. This survey is essential for optimization of routine CT protocols especially for CT units without automatic exposure control systems, which are main part of CT units in Bulgaria.

Key words: Computed tomography, dose quantities, image noise, exposure parameters, optimization.

Introduction

The manufactures of CT equipment provide standard examination protocols including exposure parameters chosen to obtain patient dose and image quality in order to satisfy the target diagnostics criterions. The choice of exposure parameters takes into account basic requirements for radiation protection and technical limitations of X-ray tubes. The large dynamic range of detectors used in CT allows wide choice of exposure parameters. These large variations of exposure parameters permit dose optimization of CT examinations without deterioration of image quality. The justified choice of basic exposure parameters is the most important factor in pediatric CT examinations [7]. Basic exposure parameters in CT are tube potential, tube current, rotation time and slice thickness. The investigation of how exposure parameters influence patient dose provides first step in patient dose reduction in CT examinations. On the other hand patient dose reduction worsens image quality.

The main aim of this paper is to study the influence of basic exposure parameters on dosimetry quantities and image noise in computed tomography. Another issue is to use received data for optimization of clinical CT protocols, which is the main task in CT because of high radiation dose delivered in CT examinations. This task is especially important for CT systems without automatic exposure control, what is the case for most of the CT systems in Bulgarian hospitals.

Material and method

The choice of exposure parameters is a complex task and depends to a large extent on the anatomical region to be scanned, the size and pathology of the patient. The chosen exposure parameters should result in sufficient image quality needed to aid clinical diagnostic. The main problem in determining exposure parameters is image noise and its effect on image quality. The main exposure parameters influencing patient dose in CT are divided in two groups: direct and indirect parameters. The firs group parameters are tube potential, tube current, rotation time, and slice thickness, width of object, pitch or table feed per rotation, focus to axis distance and usage of dose reduction techniques. On the other hand some factors which influence directly on image quality carry weight on patient dose, like reconstruction kernel. Choice of these parameters is under control by the operator and can be used for optimization of image quality and patient dose.

Energy fluence rate ψ of X-ray is the basic quantity that determinates the patient dose. ψ is influenced by electrical parameters of X-ray tube: tube potential U_a and tube current I_a, and atomic number of the target Z:

$$\psi = k \cdot U_a^n \cdot I_a \cdot Z \tag{1}$$

k is proportional coefficient and index n depends on tube filtration, (n=2÷5). It is evident from this equation that patient dose can be reduced by the choice of exposure parameters, but ensuring sufficient image quality is obtained.

Image noise in CT is determined by statistical fluctuations of Hounsfield units, evaluated in region of interest in homogeneous image. Image noise can be determinated from standard deviation of Hounsfield units, estimated in homogeneous image. CT number values are proportional to the number of registered photons N:

$$\sigma = \frac{\sqrt{N}}{N} \tag{2}$$

Image noise has three components: quantum noise, electronic noise and structural noise. The main noise source is quantum noise, defined by the statistical fluctuation of registered photons. Quantum noise depends on basic exposure parameters: tube potential, tube current, rotation time, and beam filtration, slice thickness and reconstruction kernel. Electronic noise gives small contribution to the image noise. The level of structural noise is determined from scanning geometry, system calibration and reconstruction algorithm. Structural noise is prerequisite to artifacts in the image [1]. Kernel filters have significant influence on image noise [10]. Smoothing kernel filters decrease image noise when exposure parameters are kept constant. The choice of kernel filters in CT protocol depends on clinical task. The operator can select reconstruction kernels. Optimal reconstruction kernel in conjunction with optimal exposure settings can reduce patient dose without decreasing the image quality [6].

Factors affecting directly patient dose and image noise

Scanning geometry and beam quality. Specific construction of CT units includes scanning geometry, beam quality and beam shaping filters. Beam shaping filters determinates patient dose and image noise. In some CT units there are one or more beam shaping filters. Bow-tie filters decrease patient dose up to 50% in comparison with flat filters [12]. At the same time bow-tie filters improve image noise homogeneity.

Basic exposure parameters affected on dosimetry quantities and image noise in CT under operator's control are tube potential, tube current, rotation time and slice thickness.

Tube potential U_a is not adjustable in most of the CT systems, which have only few discreet settings possible. Typical values of tube potential in computed tomography units are in the range of 120-140 kV. Higher value of tube voltage and respectively high beam energy leads to increase of patient dose and decrease image noise respectively due to the increased penetration of the beam. In newer days there is a tendency to use lower tube potential from 80 kV to 110 kV, especially for pediatric examinations [5]. The advantage of lower tube voltage is improved contrast between lesions and surrounding tissue. Reducing tube voltage from 120 kV to 80 kV, when other exposure parameters are kept constant, may lead to decreasing of effective dose up to factor of 3. Diagnostic CT images obtained with lower tube potentials may contain artifacts due to the beam hardening.

Time-current product Q is the basic parameter influencing patient dose and image quality. Q is determinates as a product of tube current I_a and rotation time t:

$$Q = I_a \cdot t, \text{ mA.s} \tag{3}$$

Image noise is defined as reciprocal of the squire root of time-current product [11]:

$$\sigma \sim \frac{1}{\sqrt{Q}} \tag{4}$$

Tube current determines fluence rate (intensity) of irradiated photons per time period. The increase of tube current leads to improvement of signal to noise ratio, due to the increase of the flux of electrons reaching the target. Decreasing the tube current is the most commonly applied patient dose reduction technique. The reduction of tube current by 50% leads to half the patient dose [13].

Rotation time determinates exposure duration, influencing both primary and scatter radiation. Most of the CT units use several fixed values of rotation time depending on scanned area. Selection of this parameter is performed according to the main diagnostic task. In theory rotation time should be as low as possible in order to minimize the motion artifacts in the images during scanning procedures due to anatomical movements. On the other hand shorter rotation times increases image noise. Therefore optimal choice of rotation time should be done according to the clinical usage of CT examination. Shorter rotation time must be used for scanning areas with higher value of anatomical noise like thorax examination. For head CT examination anatomical noise level is low, therefore longer rotation times should be used in order to achieve lower quantum noise in the image.

Slice thickness also influences patient dose and image noise. In theory decreasing of slice thickness lead to increasing of image noise due to decreasing of ψ. Image noise level is increasing with $1/\sqrt{h}$, where h is nominal slice thickness [8]. Therefore using slices with thickness of 10 mm will lead to decrease of image noise by factor of 3.2 in comparison to 1 mm slice. On the other hand thin slices ensure better spatial resolution

[3]. Selection of slice thickness should be made according to clinical task of CT examination.

In helical scanning mode patient dose increases with the decrease of pitch factor value [2]. Image noise dependency on pitch value is more complex and strongly depends on the type of CT unit: for single slice CT systems image noise is independent, while in multislice CT units image noise depends on pitch value [9].

This study is focused predominantly on adjustable by the operator CT parameters that affect radiation exposure and image noise. The study was performed on five clinical CT units, all of them of third generation – GE Prospeed Plus (Unit 1), GE BrightSpeed 16 (Unit 2), Siemens Somatom AR.C (Unit 3), Siemens Somatom Emotion (Unit 4) and Siemens Somatom Emotion Duo (Unit 5). First four CT units are single sliced and unit 5 is dual sliced system. Investigation on the influence of main exposure parameters on patient dose and image noise was performed in axial scanning mode.

Dose quantity measured for all systems was Computed Tomography Dose Index free in air on axis of rotation, $CTDI_{air}$. It was measured with pencil type ionization chamber with active length of 100 mm, type TW30009, connected to a radiation measure device Unidos E type T10-009 (PTW Freiburg, Germany). Image noise estimation was performed with measurements of the standard deviation of CT numbers in homogeny area of 500 mm^2 in the centre of image on water equivalent phantoms. Water phantom from the set of every particular scanner was used. Water phantoms used with Siemens, Shimadzu and GE units were with diameter of 200 mm, 230 mm, and 250 mm respectively. The phantoms were positioned in the center of rotation of gantry. The measurements were performed in axial scanning mode using standard protocol for head examination.

Results and discussion

The results of the study can not be used for direct comparison between scanners, due to differences in geometry design and X-ray beam quality. The obtained results were used for investigation of the influence of exposure parameters on $CTDI_{air}$ and image noise.

The dependence of $CTDI_{air}$ on tube voltage, tube current, rotation time and slice thickness are presented on Figures 1-4 respectively and the dependence of image noise on the same parameters are presented on Figures 5-8 respectively.

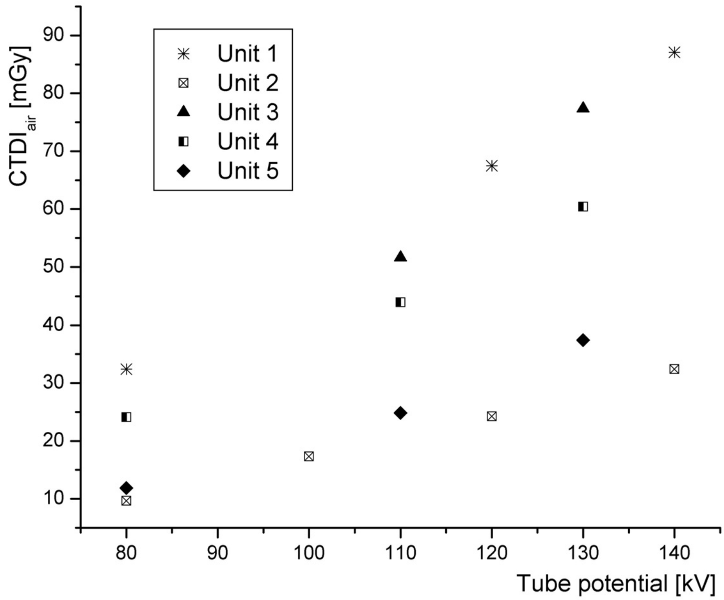

Figure 1. Dependence of $CTDI_{air}$ value on applied tube potential.

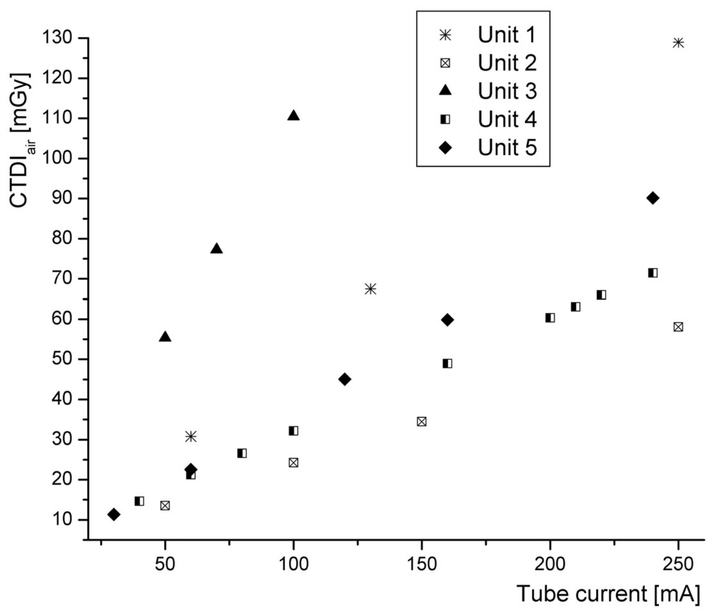

Figure 2. Dependence of $CTDI_{air}$ values on tube current.

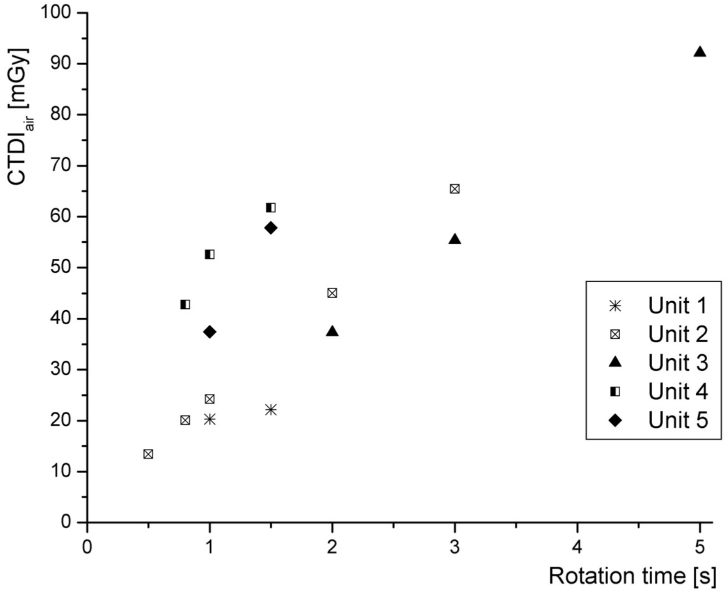

Figure 3. Dependence of $CTDI_{air}$ values on rotation time.

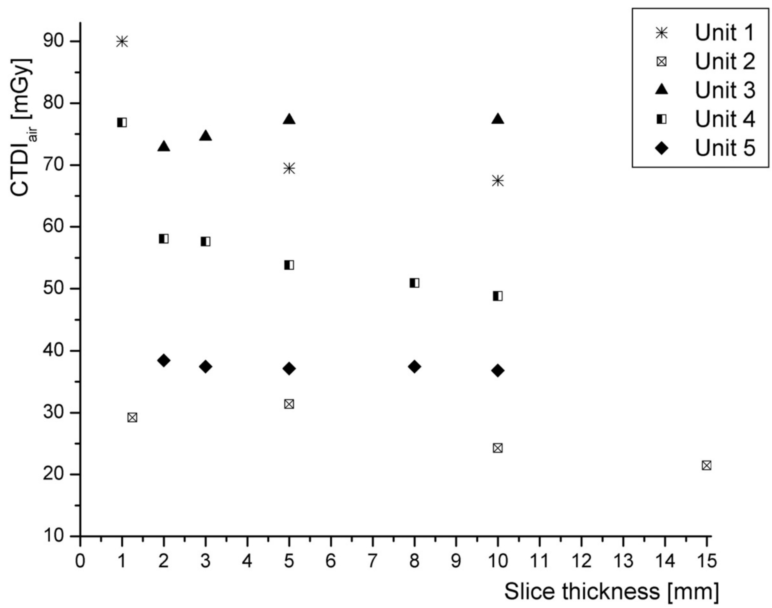

Figure 4. Dependence of $CTDI_{air}$ value on slice thickness.

Influence of main exposure parameters on dose values

The mostly used tube voltages are in the range 120-140 kV depending on the scanner type with typical value of 120 kV. In some special applications, like thorax and pediatric CT, lower tube potential 110 kV is used. It was found that even minor alteration of tube voltage at the same time keeping constant current-time product, leads to significant change of patient dose (Figure 1). This is due to the strong dependence of the X-ray beam intensity on the squire root of the tube voltage. The ddecrease of tube voltage with 20 kV, from 130 kV to 110 kV and from 140 kV to 120 kV, diminishes the dose value with 22-34% respectively. In the same time this leads to increase of image noise from 3 to 25% (Figure 5). The survey of clinical CT practice in Bulgaria showed that in most of the hospitals where CT equipment without AEC is in use, CT examinations from particular type are performed for all adult patients with the same tube current value. Therefore, for these cases optimization is possible with adopting the tube current values to the weight of the patient, which will reduce patient doses without deterioration the image quality. From Figure 2 it is evident that for all CT units the dependence of $CTDI_{air}$ value on selected current values is close to linear. It was found that unit 3 have stronger dependence than the rest of the CT systems included in this study. Reduction of tube current is the most effective practical means of reducing radiation dose in CT examinations. Any decrease in tube current should be considered, because such reduction causes an increase in image noise (Figure 5) which may affect the diagnostic outcome of the examination. Automatic exposure control system in modern CT units is based on the tube current modulation [4]. Strong dependence close to linear was found between $CTDI_{air}$ value and rotation time for all units included in the study (Figure 3). From Figure 4 it is evident that $CTDI_{air}$ depends on the slice thickness. Large variation of the dependence of $CTDI_{air}$ on slice thickness was found between units included in this study. This is due to the differences in focus to axis distance, scanner geometry and design. The largest dependence was found for small slice thicknesses due to the decrease of irradiated area. One of the systems (unit 5) demonstrated very small dependence of $CTDI_{air}$ on slice thickness.

Dependence of exposure parameters on image quality.

Figure 5 demonstrates that the decrease of tube potential when other exposure parameters are kept constant causes significant increase of image noise. Lower tube voltages used with not sufficient tube current compensation when scanning large

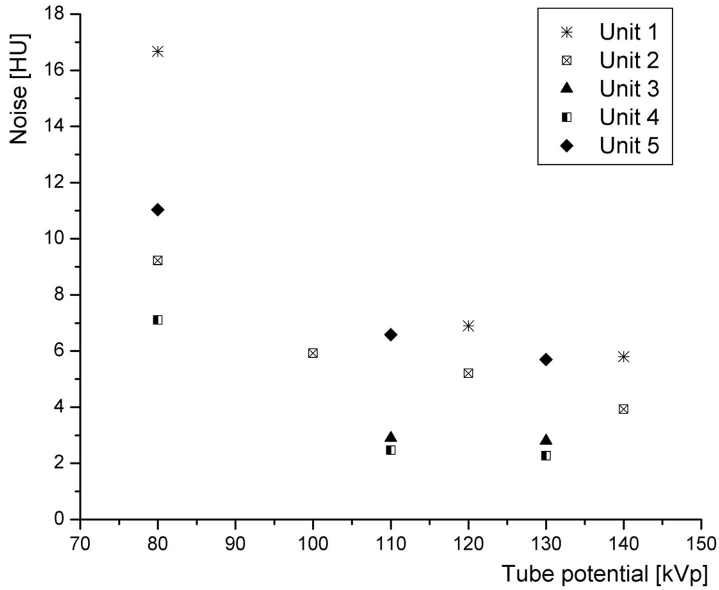

Figure 5. Image noise dependence on applied tube voltage.

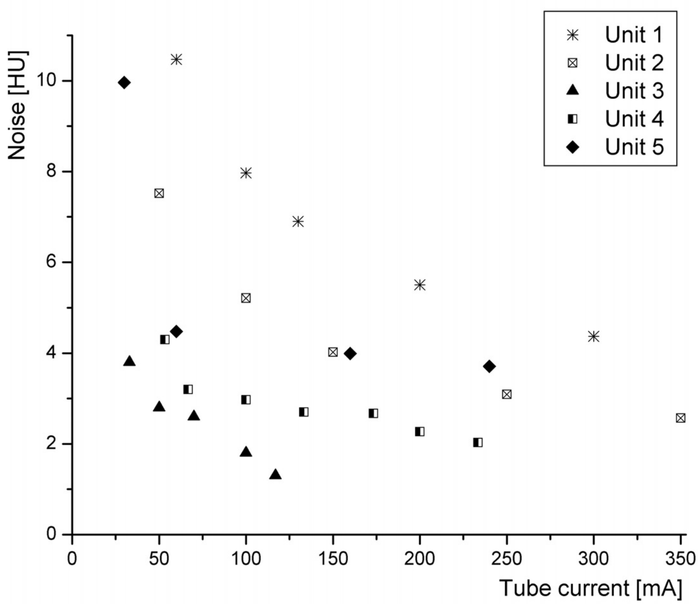

Figure 6. Dependence of image noise on tube current.

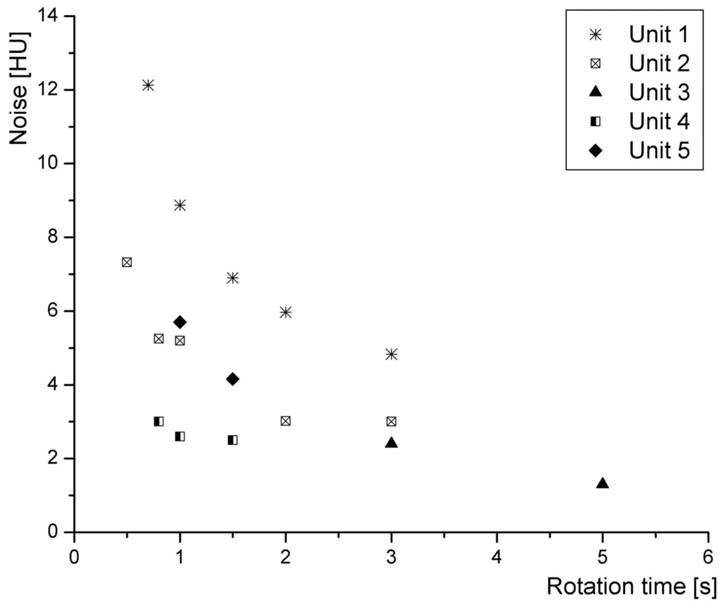

Figure 7. Dependence of image noise on rotation time.

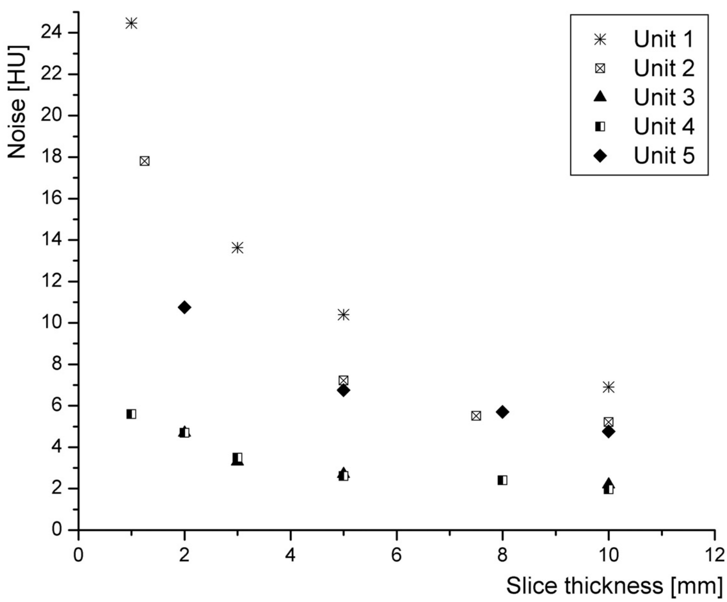

Figure 8. Dependence of image noise on slice thickness.

patients may lead to insufficient image quality, due to significant increase of image noise. It is evident from Figure 5 that typical operating ranges of tube potential in Bulgarian CT units ensure sufficient noise level in images. From Figure 6 it is evident that tube current had strong dependence on image quality. It was found that when tube current is decreased four times, the image noise increases with 90%. For large patient examinations higher tube current values are usually used in order to improve image quality. In this cases should be kept in mind that higher tube current causes increase of patient dose (Figure 2). Image noise dependence on rotation time for a single rotation is presented on Figure 7. More detailed investigation was performed only on two CT units due to technical limitations. It was found that the doubling the rotation time decreases the image noise between 50% and more than 70%. The investigation of slice thickness on image noise dependence showed that the noise is higher for narrow slices due to the decreased number of detected photons. It was found that twice thinner slice is associated with 23% to 50% higher image noise. The mostly used slice thicknesses for head and body CT examinations in our clinical practice are in the range of 5 mm to 10 mm. These values ensure lower image noise values with acceptable patient dose values (Figure 4); on the other hand wider slices decrease image spatial resolution [3]. Figure 8 demonstrates less dependence of image noise on slice thickness for units 3 and 4 in comparison to other CT units. This might be due to the differences in technical design and in filter kernels. The specific filter kernel used in standard head protocol at these two systems is the reason for twice lower image noise.

Conclusion

The studied influence of the exposure parameters on dosimetry quantities and image noise demonstrated the basic ways for dose and image noise optimization for CT units without AEC systems. Due to the existing differences in scanner design and beam qualities the results can not be used for direct comparison between different scanners.

The alteration of tube voltage and tube current show the potential in patient dose reduction up to 50% but may deteriorate image quality, and has to be carefully applied for every particular scanner.

The optimization of clinical protocols is possible but has to be done taking into account the specific clinical task.

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