CHARACTERISTICS OF X-RAY BEAM USED IN COMPUTED TOMOGRAPHY

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Introduction

Patient exposure from Computed Tomography (CT) was simulated using Monte Carlo method. The model of rotating source was implemented previously.

Conversion coefficients from measured dose indexes to doses in organs and tissues of patient were determined. Each coefficient is a quotient of calculated organ dose divided by a calculated computed tomography dose index ($CTDI_w$). Simulation using Monte Carlo method requires the energy spectrum of incident radiation. At a given anode voltage the anode angle and total beam filtration determine the energy spectrum. When these quantities are unknown, measurements of attenuation in cylinder dosimetric phantom can be used to characterize the spectrum.

A comparison of radiation doses to patients during CT examinations was performed in [1] and showed that the $CTDI_w$ measured in CT scanners from various manufacturers have large discrepancies while input settings are the same. Organ doses differed by a factor of two and the similar image quality. Hence, for accurate patient doses the calculation should be scanner-specific. At the same time variation of conversion coefficients is much smaller (several percent). This suggests using of a limited set of conversion coefficients, calculated for a number of combinations of filters and anode angles. These values can be matched to calculated coefficients based on $CTDI_w$ measurements.

Energy spectrum can be characterized using the quotient of $CTDI_w$ measured in cylinder dosimetric phantom to $CTDI_{air}$ measured free-in-air. This relation lies in the range of 0.2 to 0.77 for different CT scanners [2, 3]. Based on this relation appropriate set of conversion coefficients can be chosen to estimate organ doses. Current quality assurance protocols requires only phantom measurements of $CTDI_w$. This approach can provide doses to adults only [4].

Apart from the energy spectrum beam profile also impacts the dose. Simulation needs the shape of bow-tie filter that determines the beam profile. The beam profile can be measured using ionization chamber while the X-ray tube position is fixed. Several beam profiles were published [5] and can be used in simulation of specific scanners.

Materials and methods

Monte Carlo simulation was employed to simulate the dynamic beam of the CT. The software is based on the method developed by M. Ghita [5]. The input spectrum of energy of X-ray beam is set as a file “probabil.txt” with a single column of numbers. Each value represents relative intensity of radiation in a 1 keV bin. The following relation is well-known:

$$E_{max}=q_e U, \tag{1}$$

where E_{max} is the maximum energy in the spectrum of the X-ray tube, Joules,

q_e is the elementary charge, Coulombs,

U is the anode voltage, Volts.

However if one converts the units of E_{max} into electronVolts, then (1) can be written in even more simple form:

$$Emax=\frac{electronVolt}{Volt}U \tag{2}$$

In our case the numeric value of anode voltage is equal to number of values in the file “probabil.txt”.

TASMIP [6] semiempirical model and IPEM78 catalogue [7] were used for generation of spectra for various anode voltages, anode angles and filters.

For needs of CT dose assessment $CTDI_{air}$ (free-in-air), $CTDI_w$ (in body phantom) were studied. Also to characterise the attenuation of the X-ray beam the air kerma free-in-air was studied. The former case was simulated with a fixed position of X-ray tube in contrast with the ordinary rotation mode as it happens in CT.

The general scheme of CT beam is shown in Figure 1.

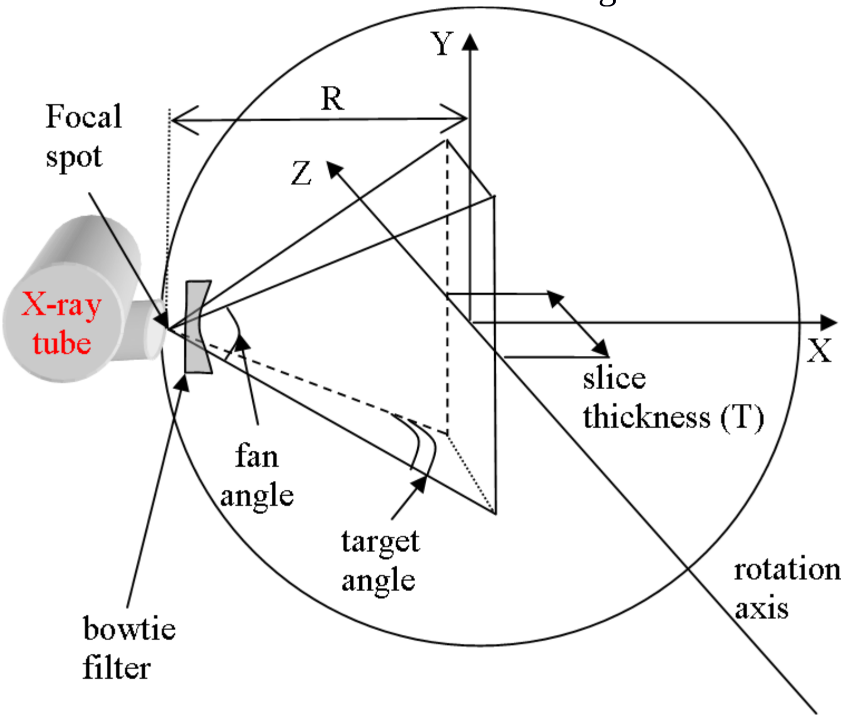

Figure 1. Dimensions of X-ray beam in a CT, R – radius of X-ray tube trajectory.

Results

Values of $CTDI_w$ were calculated in axial mode for beam width 1-4 mm (slice thickness in Figure 1). $CTDI_w$ has very weak dependence on the slice thickness. Table 1 presents $CTDI_w$ calculated in body phantom (TASMIP model, anode angle 12,5°, 100 kV, 0 mm added filtration, X-ray tube trajectory radius 60 cm, fan angle 49,2°, axial mode).

Table 1. Dependence of $CTDI_w$ on beam width.

Slice thickness, cm	$CTDI_w$, Gy/photon	Deviation from average, %
2.5	$3.305\cdot10^{-16}$	0,08

3	$3.303 \cdot 10^{-16}$	0,017
3.5	$3.304 \cdot 10^{-16}$	0,04
4	$3.298 \cdot 10^{-16}$	0,14
average	$3.302 \cdot 10^{-16}$	0

The relative error caused by Monte-Carlo method in each case is less than 0,2%. Values were calculated separately for four peripheral and one central hole of the phantom. Number of sampled photons was $5 \cdot 10^7$. Each calculation took about 30 minutes of computation time.

The dependence of $CTDI_w$ on the added filtration is shown in the Table 2. Values were calculated for anode voltage 80 kV, beam width is 3 mm Al. Other parameters are the same as for Table 1.

Table 2. Dependence of $CTDI_w$ on added Al filter thickness.

Al filter thickness d, mm Al	$CTDI_w$, Gy/mA/sec	$\frac{CTDI_w}{CTDI_w(0\,mm\,Al)}$
0	$4{,}82 \cdot 10^{-10}$	1
0,5	$4{,}21 \cdot 10^{-10}$	0,87
1,4	$3{,}41 \cdot 10^{-10}$	0,71
2,5	$2{,}78 \cdot 10^{-10}$	0,58
3	$2{,}51 \cdot 10^{-10}$	0,52
4	$2{,}12 \cdot 10^{-10}$	0,44

Errors from Monte-Carlo method in each value is less than 0,2%. Number of sampled photons varied from $3{,}5 \cdot 10^7$ to $5 \cdot 10^7$.

Discussion

Calculations show weak dependence of $CTDI_w$ on beam width. This provides an opportunity to reduce the required number of calculations for building a database of conversion coefficients for patient dose assessment from CT. On the other hand the strong dependence of $CTDI_w$ on added filter thickness. This dependence is well fitted by exponential function:

$$\frac{CTDI_w}{CTDI_w(0\,mm\,Al)} = 0.97 \cdot \exp(-0.203\,d) \tag{3}$$

Actual thickness of added filtration is usually unknown. The accurate measurement of attenuation of an X-ray beam from CT and comparison with known curve (3) gives a the necessary value for further calculations. However it is not usually possible to measure it in clinic. In this case the method developed by Turner [1] can be used.

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