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Measurement of organ dose in abdomen-pelvis CT exam as a function of mA, KV and scanner type by Monte Carlo method

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ABSTRACT

Background: CT is a diagnostic imaging modality giving higher patient dose in comparison with other radiological procedures, so the calculation of organ dose in CT exams is very important. While methods to calculate the effective dose have been established (ICRP 26 and ICRP 60), they depend heavily on the ability to estimate the dose to radiosensitive organs from the CT procedure. However, determining the radiation dose to these organs is problematic, direct measurement is not possible and comparing the dose as functions of scan protocol such as mA is very difficult. One of the most powerful tools for measuring the organ dose is Monte Carlo simulation.

Materials and Methods: Today the predominant method for assessment of organ absorbed dose is the application of conversion coefficients established by the use of Monte Carlo simulations. One of the most famous dose calculation software is CTDOSE, which we have used it for calculation of organ dose. In this work we measured the relationship between the mA, KV and scanner type with the equivalent organ dose and effective dose in mathematically standard phantom (Hermaphrodite 170cm/70Kg) in an abdomen-pelvis CT exam by Monte Carlo method. For this measurement we increased the mA in steps of 10 mA and plot curves for organ dose as a function of mA for different KV setting.

Results: As expected, with increasing mA, patient organ dose increased, but the simulation results showed that the slope of organ dose as a function of mA increased with KV increasing. By increasing KV from 120 to 140 the increase in slope of curves representing patient organ dose versus mA for different scanner types show almost similar behavior whereas the slope of the corresponding curves in scanners which equipped xenon detectors was almost 22% more than the slope of scanners equipped with scintillation detectors.

Conclusion: Our research showed that regarding equivalent dose the system incorporating scintillation detector has a superior performance. Incorporating such software in various CT scanners, marketed by different vendors, will offer the ability to get a print out of patient organ dose in any examination according to the imaging parameters used for imaging any part of the body.
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INTRODUCTION

CT Dosimetry

In X-ray diagnostics, only parts of the human body are exposed during an examination, and due to the relatively low energy of photons, the energy deposition is very inhomogeneous.

Furthermore, different kinds of tissues and organs have different sensitivities for radiation. In order to take these circumstances into account, the effective dose, ***E*,** was given by ICRP (ICRP 1996):

$$E = \sum_T w_T \cdot H_T \qquad (1)$$

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Where w_T is the tissue weighting factor, and H_T is the equivalent dose to organ or tissue *T*. The effective dose is the bottom line for statements about patient doses and the associate risks also for computed tomography (CT). However, the ways to find necessary knowledge about organ doses from CT examinations are not straight forward. A number of measurement methods have been reported that use a variety of ways to describe or characterize the radiation delivery by CT, and these are quite different from methods and procedures normally employed in conventional X-ray diagnostics. A fundamental dosimetric quantity is the computed tomography dose index, *CTDI*, defined by:

$$CTDI = \frac{1}{T}\int_{-\infty}^{+\infty} D(z)dz \qquad (2)$$

Where *T* is the thickness of the tomographic slice, and *D(z)* is the distribution of absorbed dose along a line parallel to the axis of rotation, designated the *z* axis. The *CTDI* may be assessed free in air or in phantoms, and the measurement may be done with TLDs or ionization chambers (Olerud 1999). The Food Drug Administration (FDA) in the US recommends doing the measurements in the center and periphery of cylindrical PMMA phantoms of 16 cm and 32 cm diameter, respectively. Because of the scattered radiation in the phantom, the total integration length must be defined.

According to FDA, the dose is to be integrated over 14 slices thickness, which implies that the total integration length depends on the slice thickness. This approach was adopted by the IEC in 1994, but is not very practical, so the predominant method is now to apply a fixed integration length of 100 mm for all measurements (IEC 1998).

There is, however, still some confusions concerning the definition and interpretation of various quantities found for single and multi slice CT scanner. This is basically caused by the definition of the nominal tomographic section thickness, *T*, and the number of tomographic section, *N*. As suggested by Cynthia and Zink (Cynthia *et al.* 1999), there is a need for the definition of a new quantity; the total nominal tomographic sections thickness, T_T, representing the total nominal active detector width. The $CTDI_{100}$ would then be defined as:

$$\mathrm{CTDI}_{100} = \int_{-50mm}^{+50mm} \frac{D(z)}{T_T}dz \qquad (3)$$

where *D(z)* is the dose profile along a line *z* perpendicular to the tomographic plane and T_T is the total nominal tomographic section thickness, which in turn is defined as the width of the sensitivity profile taken at the center of a tomographic section. For multi slice equipments, this width is the sum of the sensitivity profile of all active detectors.

The weighted and normalized ${}_nCTDI_w$ was defined as:

$${}_nCTDI_w = \frac{1}{C}\cdot\left(\frac{1}{3}\cdot CTDI_{100,c} + \frac{2}{3}\cdot CTDI_{100,p}\right) \qquad (4)$$

Where $CTDI_{100}$ is measured in the center and periphery (1 cm under the surface) of a 16 cm (head) and 32 cm (trunk) phantom, respectively; division normalizes the weighted quantity with the current time product per slice, *C* (mAs). The actual $CTDI_w$ is obtained by multiplying with the mAs value used in the clinic. Correspondingly, the dose length product *DLP*, in units of mGy.cm, for a complete CT examination was defined for conventional axial CT and for helical CT, respectively, as:

$$DLP_{Helical} = \sum_i {}_nCTDI_w \cdot T \cdot A \cdot t$$
$$DLP_{Axial} = \sum_i {}_nCTDI_w \cdot T \cdot N \cdot C \qquad (5)$$

Where *T* is the slice thickness, *N* the number of slice, *A* the tube current, and *t* the total acquisition time.

Relationship between Different Dose values

CTDI is a parameter that gives information about the dose due to a particular set of acquisition parameters. It is useful to allow dose comparisons between protocols for a given CT

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system, and can be used to calculate the dose length product. However, it should be used with caution when comparing patient doses from two different scanners.

DLP can be used as an indicator of the radiation dose delivered throughout the examination.

The *Effective Dose* can be derived from the *DLP* and the normalized effective dose factors for each of the organs irradiated during the examination. This parameter is the most important one in making risk benefit-decisions, because it gives an estimation of patient dose that can be related to biological risk (GE 2001).

The relationship between these parameters is shown in figures 1 and 2.

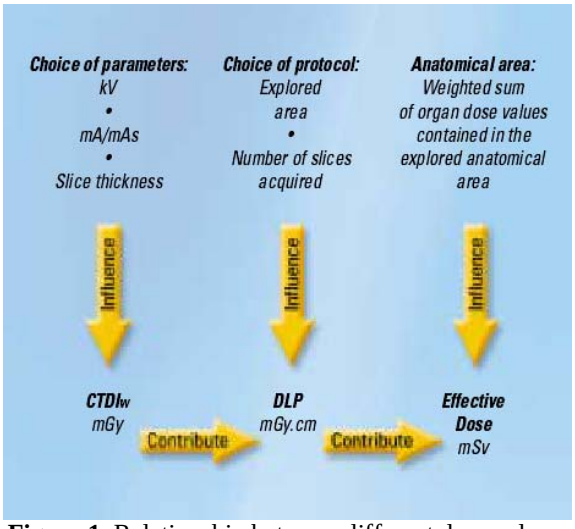


Figure 1. Relationship between different dose values.

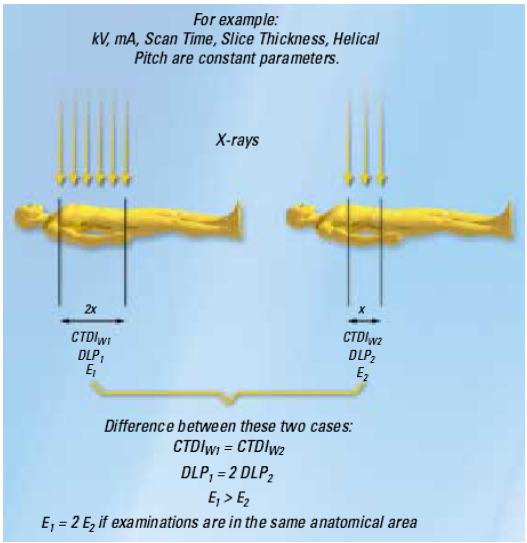


Figure 2. Relationship between dose and scan area.

Factors influencing patient dose in CT

It is often the case that an improvement in image quality is obtained at the expense of a higher radiation dose to the patient. Conversely, when reducing the dose, the quantity of X-rays reaching to the detector decreases as well. The parameters, which affected in patient dose divided in two different categories; Equipment-related factors and Application-related factors.

The Equipment-related factors are: waveform of the generator, range of tube current settings, beam filtration, beam shaper, focus-axis distance, slice collimation, detector array, scan geometry, partial fan beam scanners and scan angle.

The Application-related factors are: exposure time, object diameter, slice thickness, pitch factor, number of slices, reconstruction filter, window width, and matrix size and filed of view (Nagel 2000).

MATERIALS AND METHODS

Today the predominant method for assessment of organ absorbed dose is the application of conversion coefficients established by the use of Monte Carlo simulations. This technique is used in most fields of medical radiation physics. The Monte Carlo method, in this context, is a computational model in which physical quantities are calculated by simulating the transport of X-ray photons. Early attempts to model the shape of a human being and its internal organs in order to calculate absorbed radiation doses were made by Snyder *et al.* (1996) and Koblinger (1972).

The earliest version of Monte Carlo software for calculations of absorbed dose and effective dose in CT is CTDOSE, which has been written by Le Heron JC, from National Radiation Laboratory, Christchurch, New Zealand (Heron 1993).

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CTDOSE Software

In this work, we used commercially available Monte Carlo software CTDOSE (Heron 1993) for calculation of organ dose. The CT-Dose calculation program uses a Monte Carlo simulation routine to estimate dose distribution and consequently the effective dose in a mathematically standard Hermaphrodite phantom (170cm/70Kg) from a given CT-procedure and a given CT-scanner type, as well as the dose-length-product from the CT-procedure.

The CTDOSE used normalized organ dose data sets, together with measured values of free-in-air axial dose for particular models of scanner, and details of the clinical technique for each examination type (Jones *et al.* 1991, 1993). This software required the following input parameters: scanned volume (in terms of baseline in the phantom and number of slices), slice width, couch increment, effective mAs, kVp and CT dose index per mAs (CTDI).

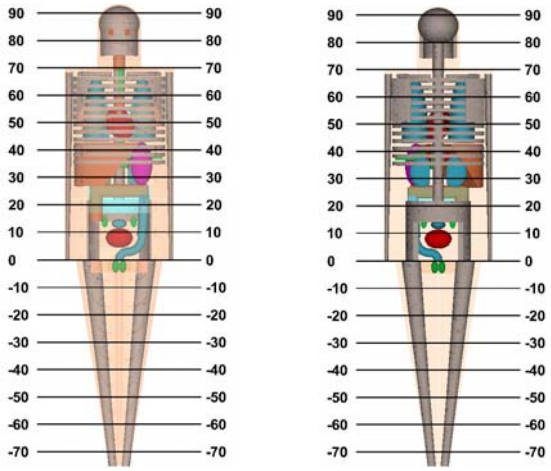


Figure 3. Hermaphrodite phantom

Measurement Conditions

One of the most popular CT exam is abdomen-pelvis scan, the routine protocol for this exam is: slice thickness 10 mm, table feed per slice 10 mm, scan start position 13.5 cm and scan end position 52.9 cm (figure 3).

For measurement of organ absorbed dose in this examination as a function of mA, we increased the mA in steps of 10 and measured the absorbed organ and effective dose for specific kVp and scanner type; on the other hand for measurement of effect of kVp and scanner type (Geometry and detection system) in organ absorbed dose we repeated increasing the mA and measured organ dose with different kVp and scanner type.

RESULTS

mA

The absorbed dose is directly proportional to the tube current (mA). The main point in this study is calculation of the slope of increasing dose via mA; this slope could be helpful to estimate the absorb dose caused by increasing mA.

Using CTDOSE Monte Carlo package we acquired the equivalent dose in different organs and effective dose in an abdomen-pelvis exam as a function of mA (30 mA up to 200 mA in steps of 10 mA) at 120 kVp in GE CTi Scanner (figures 4 and 5). The slope of increasing equivalent dose for some organs is calculated in table 1.

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kVp

Dose increased with an increase in kVp. To obtain the relation between equivalent dose and kVp, CTDOSE Monte Carlo package was used. In figures 6 and 7, the equivalent dose and effective dose was calculated as a function of kVp in 130 mA tube current in GE CTi scanner for an abdomen-pelvis exam.

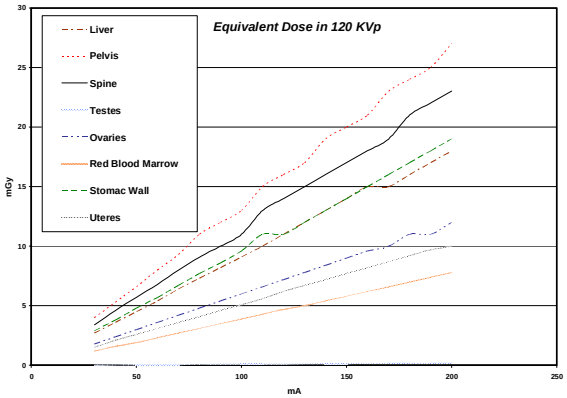


Figure 4. Equivalent dose as a function of mA.

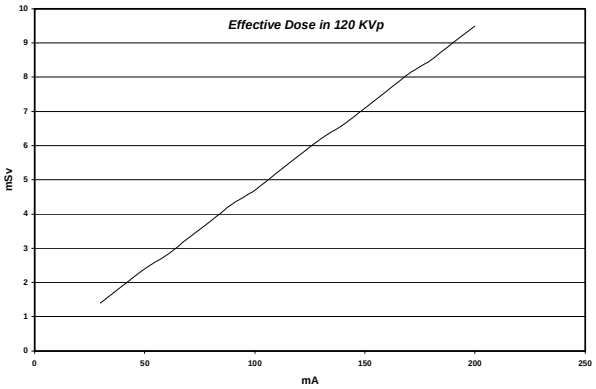


Figure 5. Effective dose as a function of mA.

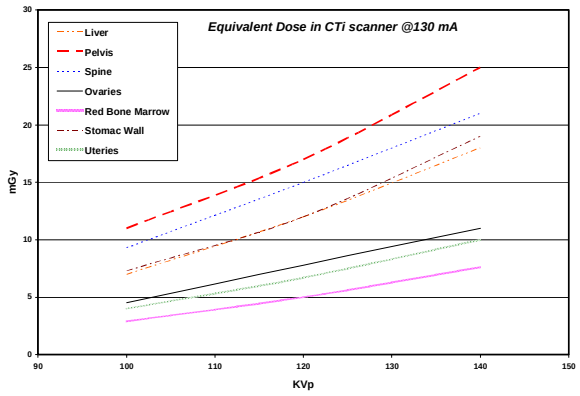


Figure 6. Equivalent dose as a function of kVp.

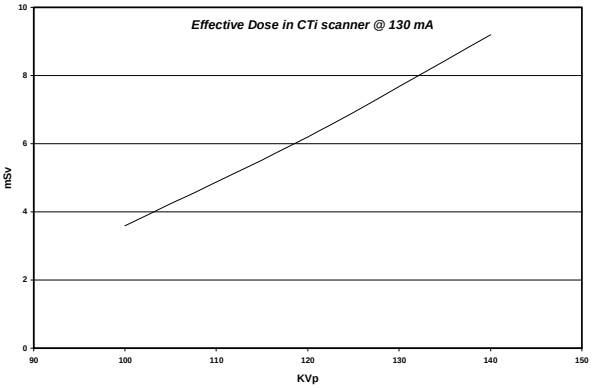


Figure 7. Effective dose as a function of kVp

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Table 1. Slope of increasing equivalent dose in terms of mA and kVp.

Organ	Slope of Increasing Equivalent Dose mGy/kVp @ 130 mA in GE CTi Scanner	Slope of Increasing Equivalent Dose mGy/mA @ 120 kVp in GE CTi Scanner
Lungs	0.18250	0.06000
Stomach Wall	0.29250	0.09470
Urinary Bladder Wall	0.03850	0.01200
Breasts	0.16250	0.06058
Liver	0.27500	0.09000
Esophagus	0.09500	0.03152
Thyroid	0.00200	0.00032
Skin	0.06750	0.02500
Bone Surface	0.13250	0.05470
Red bone marrow	0.11750	0.03882
Testes (Gonads)	0.00340	0.00090
Ovaries (Gonads)	0.16250	0.06000
LLI Wall (Colon)	0.10750	0.03411
Muscle	0.08750	0.03047
Adrenals	0.27000	0.07176
Brain	0.00001	0.00003
Small Intestine	0.26250	0.08470
ULI Wall	0.26250	0.08470
Kidneys	0.30000	0.10529
Pancreas	0.27750	0.09000
Spleen	0.26250	0.09058
Thymus	0.06000	0.01700
Uterus	0.15000	0.05000
Pelvis	0.35000	0.13529
Spine	0.29250	0.11529
Skull Cranium	0.00052	0.00004
Skull Facial	0.00232	0.00046
Rib Cage	0.37500	0.16000
Clavicles	0.01475	0.00388
Eye Lenses	0.00052	0.00001
Gall Bladder Wall	0.28750	0.10058
Heart	0.25500	0.09058

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Scanner Type

One of the most important techniques in CT dose reduction is increasing the detection efficiency. Nowadays the scintillation detectors are specifically developed for CT application. The major advantages are excellent properties such as high absorption efficiency (99%) and stability. The increase of detection efficiency allows better performance in low signal conditions such as those encountered using low dose protocols in highly attenuating patient regions.

To obtain the effect of detection efficiency in dose reduction we compared two scanner systems with different detection systems, which had fairly the same geometry (GE Prospeed with Xenon detector and GE CTi with scintillator detector). We calculated the equivalent organ and effective doses with the same techniques for both scanners.

According to our Monte Carlo simulation the effective dose for abdomen-pelvis exam in system which was equipped with scintillator detector in the same technique, (120 kVp and 130 mA) was 15% less than the Xenon detector (7.3 mSv in Prospeed and 6.2 mSv in Cti) system. Besides the equivalent organ dose in CTi scanner was less than Prospeed scanner (figure 8).

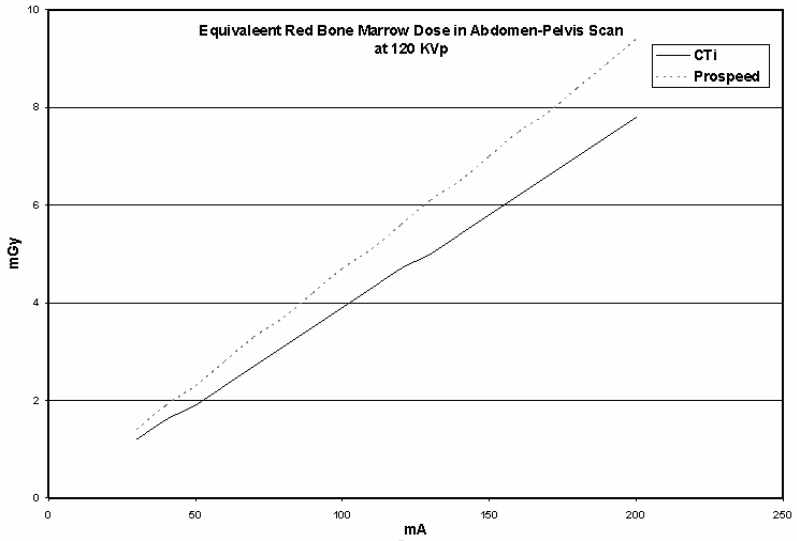


Figure 8. Equivalent dose in different Scanners

DISCUSSION

In this research we simulated the variation of equivalent and effective doses for an abdomen-pelvis exam as a function of kVp (from 100 to 140) and mA (from 30 to 200) for CTi scanner at 130 mA and 120 kVp respectively. The equivalent dose with the above mentioned parameters were simulated for different organs (figures 4 and 6). In both figures we noticed that the maximum equivalent dose and sharpest slope variation were for pelvis whereas the minimum equivalent dose with lowest slope variation was related to red bone marrow. The equivalent dose for stomach wall and liver is almost equal to each other in both figures 4 and 6.

The effective dose for the above mentioned scanner at 130 mA increased from 3.7 mSv at 100 kVp to 9.2 mSv at 140 kVp and when kVp was kept constant at 120 kVp the effective dose increased from 1.4 mSv at 30 mA to 9.5 mSv at 200 mA (Jones 1991).

Finally, the equivalent dose for red bone marrow at 120 kVp as a function of mA using two different types of scanners was simulated namely: one with Xenon detector and the other with scintillation detector (figure 8). Regarding equivalent dose the system incorporating

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scintillation detector had a superior performance (GE 2002).

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