

Patient dose considerations in computed tomography examinations

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Abstract

Ionizing radiation is extensively used in medicine and its contribution to both diagnosis and therapy is undisputable. However, the use of ionizing radiation also involves a certain risk since it may cause damage to tissues and organs and trigger carcinogenesis. Computed tomography (CT) is currently one of the major contributors to the collective population radiation dose both because it is a relatively high dose examination and an increasing number of people are subjected to CT examinations many times during their lifetime. The evolution of CT scanner technology has greatly increased the clinical applications of CT and its availability throughout the world and made it a routine rather than a specialized examination. With the modern multislice CT scanners, fast volume scanning of the whole human body within less than 1 min is now feasible. Two dimensional images of superb quality can be reconstructed in every possible plane with respect to the patient axis (e.g. axial, sagittal and coronal). Furthermore, three-dimensional images of all anatomic structures and organs can be produced with only minimal additional effort (e.g. skeleton, tracheobronchial tree, gastrointestinal system and cardio-

vascular system). All these applications, which are diagnostically valuable, also involve a significant radiation risk. Therefore, all medical professionals involved with CT, either as referring or examining medical doctors must be aware of the risks involved before they decide to prescribe or perform CT examinations. Ultimately, the final decision concerning justification for a prescribed CT examination lies upon the radiologist. In this paper, we summarize the basic information concerning the detrimental effects of ionizing radiation, as well as the CT dosimetry background. Furthermore, after a brief summary of the evolution of CT scanning, the current CT scanner technology and its special features with respect to patient doses are given in detail. Some numerical data is also given in order to comprehend the magnitude of the potential radiation risk involved in comparison with risk from exposure to natural background radiation levels.

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INTRODUCTION

Ionizing radiation can produce direct or indirect damage

to living organisms due to the ionization produced in the atoms and molecules of cells and the creation of free radicals, which may lead to both the breakage of chemical bonds and biological damage (e.g. reproductive or functional cell death, and carcinogenesis)^[1-4].

There are two kinds of effects from ionizing radiation, namely the stochastic and deterministic. Stochastic effects relate to the potential occurrence of carcinogenesis and genetic damage that may result from DNA mutations triggered by radiation. The probability of their occurrence increases linearly with dose (the current models assume zero dose threshold) and, therefore, the larger the dose the greater the risk. Carcinogenesis and hereditary effects take years to appear and their association with previous exposure to radiation is not evident. However, it has been estimated that there is a 5% probability per Sievert (Sv) for these effects to occur^[2].

Deterministic effects, on the other hand, relate to functional or reproductive death of the cells of an organ or a tissue, which may lead to injuries and functional impairment of the organ or tissue. However, this will occur only if the number of damaged cells exceeds a certain level beyond which the deficit in the cell population can no longer be compensated by normal cell reproduction mechanisms. Deterministic effects manifest themselves quite quickly (within hours or weeks) but they do have a dose threshold (> 2 Gy) under which no direct damage occurs^[4]. Typical examples of deterministic effects are skin injuries (ranging from a simple early erythema to secondary ulceration), infertility and cataract formation.

Ionizing radiation is everywhere; e.g. cosmic X-rays, radon and γ -rays that emanate from the soil and building materials, natural radioactive isotopes within our food and drink, and this comprises what is called "natural background radiation". Additionally, the population is exposed to man-made sources of ionizing radiation (e.g. medical X-rays and isotopes used in nuclear medicine, discharges from nuclear power plant normal operations and accidents, fallout from nuclear weapons testing and atomic bombs). The annual dose per caput in the UK is, on average, about 2.4 mSv and about 85% (about 2 mSv) is due to natural sources of ionizing radiation. The remaining is almost entirely due to medical applications of ionizing radiations, since their contribution to the overall annual dose has been estimated to be about 14%^[5].

Ionizing radiations are currently used in radiology, nuclear medicine and radiation therapy. Patient doses in radiology are about 10 000 times smaller than doses delivered in radiation therapy (doses delivered to tumors are on the order of 70 Gy), however, because of the vastly greater frequency of diagnostic examinations compared to radiation therapy and nuclear medicine procedures, radiology is the undisputed major contributor to the population dose from medical applications since its contribution exceeds 95%.

A few years ago, general radiological examinations (radiographs) were responsible for the largest segment of the overall contribution of diagnostic radiology to the collective population dose. However, in recent years,

the situation seems to have changed. In an old survey carried out in 1989 in the UK, it was seen that computed tomography (CT) examinations accounted for only 2% of all examinations, but yet contributed about 20% of the collective dose to the population from diagnostic imaging^[6]. About 10 years later, it was estimated that this contribution had risen to 40%-47%^[7,8] while according to a study from USA, there are hospitals where this figure has exceeded 50% for a long time^[9]. The main reason is the rapid increase in CT examinations performed, which came as a consequence of the rapid evolution of CT scanner technology.

THE EVOLUTION OF CT TECHNOLOGY AND THE INTRODUCTION OF NEW CLINICAL APPLICATIONS

In the first experimental stages of CT scanners, a simple examination could last for hours. With the improvement of computer and CT scanner technology, the commercial axial scanners could perform a chest-abdomen-pelvis examination in less than 30 min while, with the advent of helical scanners, this time was reduced to half. Then, around the end of the millennium, the multislice (or otherwise called multidetector) CT scanners came into use^[10-12]. With the latest multislice CT (MSCT) scanners, a whole body examination can be performed in less than 5 min, with most of the time spent in patient positioning rather than the actual scanning procedure, which lasts less than 1 min.

MSCT are the evolution of helical single slice CT (SSCT) scanners, which had succeeded axial CT scanners. With axial CT scanners, one image was acquired per tube rotation and, subsequently, the CT table had to be moved to the next anatomical position to proceed to the next image acquisition. Because of the time spent between scans to move the table, and the slow processing speed of the microprocessors used in the 1980s, the examination time was long compared to the current standards. The advent of helical scanners (now referred to as SSCT) speeded up the examination procedure because of the simultaneous tube rotation and table movement and also faster microprocessors and evolved software made the image reconstruction process faster.

Apart from the time gained, helical CT opened up a new world of applications ranging from the simple multiplanar reconstruction (MPR) images to CT angiography and CT endoscopy applications^[10], as well as the CT fluoroscopy that was used for real time imaging of patient anatomy during interventional procedures^[13,14]. These new applications, however, were highly demanding and CT technology was forced to go one step beyond.

Early SSCT scanners had a significant drawback: only one slice could be acquired per rotation. Thus, scanning an anatomical range of, for example, 30 cm in length using a 10 mm collimation required 30 tube rotations, when a pitch factor of 1 was used (for definition of pitch see next section), which is exactly the same number of rotations

required with axial scanning. Assuming that a tube load of 200 mAs per tube rotation was used, this amounted to a cumulative tube load of 6000 mAs, which normally could be easily accommodated by the X-ray tubes used in SSCT scanners and, thus, the examination could be performed using a single scan. However, if good quality MPR images were required, then a pitch factor of about 0.5 would be used and almost 60 rotations would be required to scan the same length. The cumulative tube load would be doubled. If these slices had to be acquired with 5 mm slice thickness instead of 10 mm (in order to increase the Z-axis spatial resolution) this would result to another doubling of the tube load. In such cases, the examination could no longer be performed with a single scan because of the thermal capacity constraints of the X-ray tubes. Additionally, some of the new clinical applications for CT, such as CT angiography, required repeated acquisition of scans before and after contrast administration. However, since in all new applications (e.g. MPR, CT angiography and CT endoscopy) the anatomic area of interest had to be covered with a single scan, it is evident that compromises were necessary (e.g. reduced mAs and scanning length, increase of pitch and slice thickness) in order to overcome the thermal capacity constraints of the X-ray tubes. Therefore, the result was not always the desired one in terms of image quality.

These problems were solved with the introduction of MSCT. Starting with the introduction of the dual slice (which is considered by some as a hybrid stage between SSCT and MSCT) and the quad slice scanners, the number of detector rows increased gradually to 6, 16, 64 and 256 with the largest number available today being 320. In parallel to the increase of detector rows, the introduction of the z-flying focal spot technology was also used by some manufacturers to convert 64-slice scanners to 128-slice scanners.

The basic differences of MSCT scanners from their predecessors, is that the MSCT scanners employ a matrix of detector elements (this is why they are alternatively called multidetector CT scanners or MDCT, in short) and a wider X-ray beam to accommodate the detector array's larger width^[10-12]. These features allow the acquisition of multiple helices of data during a single rotation; that is, multiple images in a single rotation, and consequently fast volume scanning of large regions of human anatomy in a matter of just a few seconds. However, they also resulted in an increase of patient dose, as is explained in the following sections.

QUANTIFICATION OF DOSE IN CT: BASIC DEFINITIONS

Definitions of CT dose index

The CT dose index (CTDI) was first introduced in the era of SSCT scanners^[15]. It was defined as the integral of the dose profile, $D(z)$, from a single axial scan along a line perpendicular to the tomographic plane (z-axis) divided by the nominal slice thickness (T):

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

For the case of MSCT scanners, where N slices of thickness T are acquired during a single axial scan, the following equation is used^[16-18]:

$$CTDI_{100} = \frac{1}{NT} \int_{-50mm}^{+50mm} D(z) dz \tag{2}$$

$CTDI_{100}$ is measured using a pencil type ionization chamber with an active length of 100 mm, both in free air and within two cylindrical polymethylacrylate phantoms of 16 cm and 32 cm diameter, simulating the head and body of a patient, respectively. $CTDI_{100}$ measured with the ionization chamber positioned in free air at the centre of rotation is referred to as $CTDI_{air}$. $CTDI_c$ and $CTDI_p$ are defined respectively as the $CTDI_{100}$ values measured with the ionization chamber within the centre and four positions (12 o'clock, 3 o'clock, 6 o'clock and 9 o'clock) in the periphery (1 cm for the surface) of the head and body phantoms, which are centrally positioned within the gantry^[12,16-18]. All CTDI quantities are given in units of mGy.

Definition of weighted CTDI

The weighted CTDI ($CTDI_w$) is used for approximating the average dose over a single slice and is defined by the following equation, separately for the head and the body phantoms^[12,16-18]:

$$CTDI_w = \frac{1}{3}CTDI_c + \frac{2}{3}CTDI_p \tag{3}$$

where for $CTDI_p$, the average of the four roughly equal $CTDI_p$ values measured in the periphery of the phantom is used.

Definition of volume weighted CTDI

The volume weighted CTDI ($CTDI_{vol}$) is used to account for helical scanning and is defined by the following equation^[12,18]:

$$CTDI_{vol} = CTDI_w \cdot \frac{NT}{I} = \frac{CTDI_w}{\text{Pitch factor}} \tag{4}$$

where NT is the total nominal collimation width and I is the table travel per rotation during a helical scan (pitch factor = I/NT).

Definition of dose length product

Dose length product (DLP) is used to calculate the dose for a series of scans or a complete examination and is defined by the following equation^[12,18]:

$$DLP = \sum_{i=1}^N (CTDI_{vol})_i \cdot L_i \tag{5}$$

where i represents each of the individual scans of the examination that covers a length, L_i , of patient anatomy. DLP is given in units of mGy cm.

Definition of effective dose

Effective dose (E) is a quantity that has been introduced

to quantify the biological detriment resulting from a partial body irradiation, enabling the calculation of radiological risk^[2]. Its calculation is based on the application of tissue-weighting factors (W_T) on the equivalent doses (H_T) absorbed by the various radiosensitive organs of the human body. That is:

$$E = \sum W_T H_T \quad (6)$$

$$H_T = \sum W_R D_R \quad (7)$$

For X-rays, equivalent dose and dose are arithmetically equal since the radiation type weighting factor, W_R , is equal to 1. The only difference is that equivalent dose is given in units of Sv, whereas dose is given in units of Gy.

In the case of CT scanners, E is calculated using the following equation^[16]:

$$E = \sum_{i=1}^N k_i DLP_i \quad (8)$$

where i represents each of the individual scans of the examination that result to DLP_i and k_i is a conversion factor used to translate DLP_i (in mGycm) to E (in mSv), having thus units of $mSv (mGy cm)^{-1}$. The k value depends on the anatomic region scanned and the examination type and approximate values of k for various examinations have been proposed, irrespective of the CT scanner used^[16]. Alternatively, E and individual organ doses can be calculated using software tools, such as the ImPACT CT Patient Dosimetry Calculator^[19], in conjunction with the NRPB-R250 conversion factor data set that have been derived from Monte Carlo simulations. It must also be noted that the conversion factors also depend on age, since the region of the head and the trunk of a newborn are 5 and 3 times larger, respectively, than those of an adult^[12,19-21].

It must be emphasized, however, that the actual radiation risk may be larger or, more probably, lower than suggested by the E values. First of all, the biological effects of low radiation doses have been assumed by extrapolation of data valid for high doses, assuming a zero dose threshold for stochastic effects. Furthermore, conversion factors that translate measured radiation quantities to E have been derived assuming specific geometry conditions and, most importantly, mathematical or anthropomorphic phantoms that simulate a patient of a specific body size and, therefore, their application on patients of different sizes introduces additional uncertainties.

MSCT VS SSCT: DOES MSCT INCREASE PATIENT DOSE?

The advent of MSCT scanners created serious concerns about the potential increase in radiation doses to populations and individual patients. Concerns about population dose were fully justified since new diagnostic and

interventional procedures were introduced into routine clinical practice and consequently the CT examination frequency was increased. Concerning individual patients, initial surveys did exhibit an increase in patient dose by about 34%^[10,12,22,23]. This increase was rather expected considering design differences and the fact that, with MSCT, the scanning of a large volume of the body can be easily carried out using a smaller slice width than used with SSCT^[10,12].

In MSCT scanners, the actual width of the X-ray beam was made larger than the nominal value to ensure that the penumbra lies beyond the detector active area, thus all detectors are irradiated uniformly, which is something that was not required for SSCT scanners. This resulted in a reduction of the z-axis geometric efficiency (the percentage of the X-ray beam width in the z-direction that is "seen" by the detectors), which is more prominent for smaller slice widths^[11].

In both SSCT and MSCT scanners, the use of smaller slice widths requires an increase in exposure factors to counterbalance the increased image noise. SSCT smaller slice widths had to be used with caution (because of the aforementioned limitations in the anode and tube housing assembly thermal capacity), whereas with MSCT, in which multiple images can be acquired with the same tube load used in SSCT to acquire a single image, thermal capacity limitations are relaxed and therefore overuse of smaller slice widths is more probable^[10,12].

Another factor that contributed to the increase of patient doses is something that is referred to as overscan or overranging^[10,12,24-26]. The image reconstruction algorithms used in helical tomography require the acquisition of data on both sides of the planned scan length. Thus, a number of additional rotations are performed on both sides of the scanned anatomy, increasing the total scanned length with respect to the nominal value. In most SSCT scanners only one additional rotation was performed (half one each side), whereas in MSCT up to 4 additional rotations are performed (depending on the CT manufacturer and the pitch selection)^[26]. In addition, the nominal beam width in MSCT scanners typically ranges from 2.4 to 4 cm (with the actual beam width being up to 25% larger), whereas in SSCT, the typical maximum beam width is 1 cm. Therefore, it is evident that the overscan contribution is largely increased with MSCT scanners, especially when small scan lengths are scanned. Due to overscanning, patient doses are increased since there are organs and tissues that are not imaged but are exposed as if they were within the anatomic region of interest^[25]. The problem with overscanning has been acknowledged by all major CT manufactures and in some models special collimation devices have been added to reduce the surplus patient dose^[27].

Finally, it should be mentioned that the main concerns about patient dose in CT relate to the stochastic effects of radiation (carcinogenesis and hereditary effects). However, since CT is also used for interventional procedures (such as CT guided radiofrequency ablation,

Table 1 Selected data from the UK survey on the computed tomography national reference doses

Quantity Examination/CT scanner type	CTDI _{vol} (mGy)		DLP (mGy cm)		E (mSv)	
	SSCT	MSCT	SSCT	MSCT	SSCT	MSCT
Routine brain (acute stroke)	60	82.5	760	930	1.7	2.1
Chest (lung cancer: known, suspected or metastasis)	10.5	13.5	430	580	7.3	9.9
Abdomen (liver metastasis)	13	14	460	470	6.9	7.1
Abdomen and pelvis (abscess)	13	14	510	560	8.7	9.5
Chest, abdomen and pelvis (lymphoma staging or follow up)	11	13	760	940	12.9	16

National reference doses (DRL) for computed tomography (CT) on adult patients from the 2003 review of the UK survey. For examinations consisting of two parts (e.g. the routine head CT) the tabulated volume weighted CT dose index (CTDI_{vol}) value is the average of the respective values of each stage. The DRL values have been derived as rounded third quartiles of the observed distributions using data from a large number of CT facilities in the UK. DLP: Dose length product; E: Effective dose; SSCT: Single slice helical CT; MSCT: Multislice CT.

biopsy and drainage) where the same body region may be exposed many times during a single procedure and a patient may undergo multiple procedures within a limited time scale, deterministic effects are not completely out of the picture^[14,28,29]. While the lower dose threshold for the deterministic effects of the skin is considered to be 2 Gy, stochastic effects have been also observed for skin doses just above 1 Gy^[30]. Such doses are not used very often but may occur in complicated CT guided interventional procedures^[14,28].

ESTIMATING THE DOSE AND THE RADIATION RISK

In the literature, there are many papers on patient doses in CT reporting results of small or large scale surveys carried out in CT facilities throughout the world^[6,12,22,23,31-35]. When comparing the DLP values given in these reports for typical CT examinations, it can be seen that large differences exist among CT facilities. Golding *et al.*^[36] noted that due to the large variations that exist in CT practice the differences in effective dose for a given examination may vary up to a factor of 40 between different CT facilities for a UK patient^[6] and by a factor of 20 in Norway^[37]. This is indicative of the potential danger of systematic patient overexposure in certain CT facilities.

Table 1 shows some of the results of a large scale survey carried out in the UK^[12]. The tabulated figures are the diagnostic reference levels (DRL) that have been derived as rounded third quartiles of the observed distributions using data from more than 120 SSCT and MSCT scanners. According to the definition of DRL, for 75% of the CT facilities that participated in the specific survey, the CTDI_{vol}, DLP and E mean values were smaller than the respective DRL values. Therefore, the remaining 25% of the CT facilities for which the DRL values were exceeded are considered to deliver unjustifiable high doses to examined patients. Therefore, in these CT facilities, the equipment and the techniques used should be thoroughly investigated in order to determine why they used higher doses than other CT facilities to perform the same CT examination and the appropriate corrective actions should be applied in order to reduce doses below the DRLs.

For this reason, in each CT facility a local survey should be carried out using data from a sample of 10 or more routine examinations of each examination type in order to calculate the local mean CTDI_{vol}, DLP and E values, and these should be compared with the national or international DRL in order to determine whether they are within “limits”. The quotation marks were used in the word “limits” in order to stress that the DRLs should not be used neither as a strict limit for individual examinations nor as a proof of an optimized performance. They just serve as reference values to point out cases where a systematic problem with patient overexposure may exist.

In order to appreciate the risk associated with the E values tabulated in Table 1, an example will be described: accepting the assumption of 5% probability per Sv for the occurrence of carcinogenesis or hereditary effects is valid, this means that an examination that results in an effective dose of 10 mSv (which is similar to the DRL for chest CT and abdomen-pelvis CT) involves a 0.05% probability for the occurrence of such affects. Equivalently, for every 10000 CT examinations performed (of 10 mSv each) approximately 5 individuals may be expected to develop a fatal cancer or hereditary effects as a result of the radiation exposure^[38].

While this figure may seem alarming, it must be taken into account that there is a probability of about 20% in the general population of spontaneous (unknown etiology) cancer incidence. Thus, if an individual patient undergoes 20 CT examinations of 10 mSv each during his/her lifetime, the cancer incidence probability for this individual may simply increase from 20% to 21%. However, it must be stressed again that while the current prevailing perception is that stochastic effects may be triggered even by small doses, the actual risk is uncertain due to the lack of actual data from low exposures.

Another simple way to appreciate the significance of these figures is to compare them with the annual dose per caput from natural background radiation, which is about 2 mSv. The E DRL for an abdomen CT is equivalent to the E from exposure to the natural background radiation for a period of about 3.5 years and, therefore, both involve the same risk.

From the above comparisons, it becomes evident that the use of CT may be relatively harmless in view of

the expected benefits that a correct and timely diagnosis brings. However, this statement is valid only for the use and not for the overuse of CT examinations. If CT examinations are requested and performed without proper consideration, this will definitely increase the radiation burden of the general population with unpredictable results. The continuing growth of CT use and the resulting patient doses has received a lot of attention by the medical community during recent years because it has been reported that 1.5%-2% of cancers may eventually be caused by radiation doses from CT examinations^[39].

In this context, it should be noted that an important aspect of patient radiation protection is the justification of the requested examinations. A number of documents have been published containing information on the clinical indications for CT in order to assist referring physicians to come to an informed decision on whether they should prescribe a CT exam, in the case that they may be in doubt^[40-42]. While CT is a valuable tool within the diagnostic chain, there are many cases where a CT examination is not actually necessary^[43]. Alternative radiation-free imaging modalities, such as ultrasound and MRI, should be used instead, when they may provide the same or even better diagnostic information than CT^[44].

CAN WE REDUCE PATIENT DOSES?

All modern MSCT scanners present dose saving features, such as automatic exposure control systems (i.e. automatic X-ray tube current modulation), that have been proven to substantially reduce the dose without detriment to the diagnostic quality of the CT images when properly used^[10,11,45,46]. While different CT manufacturers use different names to describe their automatic exposure control systems and differences in the design and philosophy of these systems also exist, the basic concept is the same. The information gathered concerning the attenuation properties of the scanned anatomy during the AP and/or LAT scout scans (also referred to as scanogram or topogram) is utilized to optimize the tube current during the scan so that all reconstructed images are approximately of the same quality. Most systems offer a number of preset options concerning the tube modulation mode that can be used. Depending on this selection, high mA values can be used in order to minimize the image noise, or low mA values to minimize the patient dose (with a penalty of increased image noise), or medium mA values that offer a compromise between image noise and patient dose.

Furthermore, modern CT scanners present dose awareness features, since they offer a direct display of dose information, such as the CTDI_{vol} and the DLP^[46]. This feature could potentially contribute to the reduction of patient doses, since the DLP values are readily available prior to the actual scanning and, thus, the examination protocols can be appropriately adjusted in order to offer the required image quality with the smallest possible patient dose.

It is very important to emphasize that overexposure in CT improves image quality since image noise is reduced

and, therefore, the CT images “look really nice”^[11,18,46]. However, images with less noise do not necessarily contain more diagnostic information^[47,48]. While the referring medical doctors from various specialties may prefer looking at “nice” images, it is important for radiologists to avoid this pitfall. It is their responsibility to determine the level of noise that is acceptable for each particular diagnostic task and, in cooperation with radiation technologists, to accordingly adjust the respective examination protocol parameters, bearing in mind that the patient doses should also conform to the national or international DRL. For this purpose, radiation technologists and radiologists should understand the basic CT dosimetric quantities, and work together with the medical physicists responsible for the observance of radiation protection norms in each CT facility, in order to adopt an appropriate policy concerning the technical optimization of the CT examinations performed.

CONCLUSION

The development of modern MSCT scanners has been the basis for an increase in the number of clinical applications of CT for the diagnosis and therapy of many pathological conditions. However, MSCT scanners, if not used properly, may deliver high doses to patients and, therefore, constitute a potentially significant radiation hazard. Both clinical justification and technical optimization are important to maintain a high benefit/risk ratio.

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