An Overview of Measuring and Modelling Dose and Risk from Ionising Radiation for Medical Exposures

Abstract

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BACKGROUND: The optimization of radiation dose is a legal requirement in medical exposures. This review paper aims to provide the reader with knowledge of dose by providing definitions and concepts of absorbed, effective and equivalent dose. Criticisms of the use of effective dose to infer the risk of an exposure to an individual will be discussed and an alternative approach considering the lifetime risks of cancer incidence will be considered.

Prior to any dose or risk calculation, data concerning the dose absorbed by the patient needs to be collected. This paper will describe and discuss the main concepts and methods that can be utilised by a researcher in dose assessments. Concepts behind figures generated by imaging equipment such as dose-area-product, computed tomography dose index, dose length product and their use in effective dose calculations will be discussed. Processes, advantages and disadvantages in the simulation of exposures using the Monte Carlo method and direct measurement using digital dosimeters or thermoluminescent dosimeters will be considered.

Beyond this special issue it is proposed that this paper could serve as a teaching or CPD tool for personnel working or studying medical imaging. Abstract

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Introduction

Within this special issue of Radiography several articles use Monte Carlo mathematical methods to estimate radiation dose to humans. It is appreciated that readers may have a limited knowledge of these methods, or indeed measurement methods which are used to estimate dose. For readers with limited knowledge in dose estimation this article explains a range of approaches which might be used. This article also outlines concepts and defines terms associated with dose and it discusses how data can be used to provide an indication of risk to an individual.

Ionising radiation is made up of sub-atomic particles or, in the case of X and gamma rays, it comprises electromagnetic waves from the high energy part of the electromagnetic spectrum. At energies associated with medical imaging, these particles and waves have sufficient energy to ionise an atom and liberate an electron. This process may lead to tissue damage which can result in cell mutation or apoptosis. The higher the dose of radiation, the greater chance that tissue damage will occur [1]. This probability model of biological damage is referred to as the stochastic effect. It suggests that no dose of radiation is safe. It is this worst case scenario that radiation protection is based on, in that Operators should aim to minimise the probability of tissue damage by using the least practicable amount of ionising radiation[2].

In medical imaging a range of professionals are responsible for ensuring doses are As Low As Reasonably Practicable (ALARP). The Operator performing the exposure is required to have an understanding of the steps that they can take to optimise dose thus minimising the chance of stochastic affects.

Absorbed Dose

Interactions of ionising radiation with matter can result in a proportion of radiation energy being deposited. The amount of energy deposited per unit mass is the absorbed dose (represented by the letter *D*) and is defined as joules per kilogram (Jkg^{-1}). The SI unit of absorbed is the Gray (Gy). Quantities of absorbed dose are usually quoted as milli-Gray (mGy, 1/1 000 of a Gray) or a micro-Gray (μGy, 1/1 000 000 of a Gray).

Equivalent Dose

The chance of tissue damage occurring does not just depend on the absorbed dose but also the type and energy of the radiation. Equivalent dose (represented by the symbol *H*) takes these factors into consideration and is obtained by applying a radiation weighting factor (*W*) to the absorbed dose. Radiation weighting factors are published by The International Commission On Radiation Protection (ICRP) [1]; they reflect the biological damage potential of different radiation types (Table 1). It can be considered a less fundamental quantity than absorbed dose but it is useful for indicating the health risk of radiation exposure. Equivalent dose is still defined as joules per kilogram, but is assigned the SI unit Sievert (Sv). Figures are often quoted as milli-Sieverts (mSv) or micro-Sieverts (μSv).

The equation for equivalent dose is defined in Figure 1.

Table 1 Recommended radiation weighting factors from ICRP 103[1]

Radiation Type	Radiation weighting factor
Photons (X-ray and gamma ray)	1
Electrons	1
Alpha particles	20
Protons	2

Figure 1 Equation for calculating equivalent dose from absorbed dose. [1]

$$H_T = W_R \cdot D_{T,R}$$

Where H_T is the equivalent dose
W_R is the radiation weighting factor obtained from
$D_{T,R}$ is the absorbed dose is tissue (T) by radiation type (R).

Ionising radiation that forms part of the electromagnetic spectrum (e.g. X and gamma radiation) ionise atoms through the photoelectric absorption and the Compton Effect. Both these interactions will eject an electron from an atom; this electron may ionise many more atoms. Since most of the affected atoms are ionised indirectly by the secondary electrons, photons are considered to be indirectly ionising.

Using the data in Table 1 it can be seen that an absorbed dose of 1 mGy of X-ray photons results in an equivalent dose of 1 mSv, i.e. 1 mGy x 1 (radiation weighting factor for X-ray photons), where 1 mGy of alpha particles results in an equivalent dose of 20 mSv i.e. 1 mGy x 20 (radiation weighting factor for alpha particles). In other words alpha particles have a higher risk to biological tissue.

Effective Dose

Effective dose (represented by the symbol *E*) takes into account the type and amount of exposed tissue. Different tissues within the body have difference sensitivities to radiation meaning a dose applied to one area of the body can carry a higher risk than the same dose applied to another. Effective dose takes the equivalent doses of a number of organs and through the application of a tissue weighting factor, the sum of these aims to provide a single number that is proportional to the detriment from a particular exposure. It allows comparisons of the risks associated with different imaging techniques or modalities.

The tissue weighting factors (Table 2) represent the sensitivity of their respective tissue, for example bone marrow is highly sensitive to radiation and so has a weighting factor of 0.12 where the brain is less sensitive and so has a weighting factor of 0.01. The sum of the tissue weighting factors is 1 and so the sum of the weighted equivalent doses would provide a whole body effective dose. Performing this process for different techniques allows for a comparison of doses and an indication of the detriment of these.

Table 2 Tissue weighing factors from ICRP 103 [1]

Organ	Tissue Weighting Factor
Gonads	0.08
Bone marrow	0.12
Colon	0.12
Lung	0.12
Stomach	0.12
Breast	0.12
Bladder	0.04
Liver	0.04
Oesophagus	0.04
Thyroid	0.04
Skin	0.01
Bone (surface)	0.01
Salivary Glands	0.01
Brain	0.01
Remainder*	0.12
TOTAL	**1.00**
*Remainder tissues: Adrenals, Extrathoracic (ET) region, Gall bladder, Heart, Kidneys, Lymphatic nodes, Muscle, Oral mucosa, Pancreas, Prostate (♂), Small intestine, Spleen, Thymus, Uterus/cervix (♀)	

Effective dose is still defined as joules per kilogram and also has the same SI unit as equivalent dose, Sievert (Sv, with figures quoted as milli-Sieverts (mSv) or micro-Sieverts (μSv)). The equation for calculating effective dose is shown in Figure 2.

Figure 2 Equation for the calculation of effective dose. [1]

$$E = \sum_T W_T \cdot H_T$$

Where E is the effective dose to the entire body

W_T is the tissue weighting factor of tissue (T) defined by ICRP 103

H_T is the equivalent dose absorbed by tissue (T)

Controversies with Effective dose

Effective dose is commonly used in medical imaging to compare the risks from different modalities (for example, CT of the cervical spine versus conventional radiographic imaging of the same anatomical area) or examinations that have differing dose distributions (for example, comparison of effective dose from an antero-posterior hip to that from an antero-posterior shoulder). The application of the tissue weighting factors to the equivalent doses of the organs provides the whole body effective dose from that exposure. The application of effective dose is useful in these situations as it provides Referrers, Practitioners and Operators with data that allows them to make decisions during the referral, justification and optimisation of medical imaging procedures [2, 3]. When used for this purpose effective dose is a useful figure to use, however a number of publications use this figure to calculate the risk of the exposure to an individual. As noted by a number of authors, and the ICRP themselves, the effective dose concept is not intended to be used this way as a number of factors are not taken into consideration [3-8].

The tissue weighting factors are averaged over all ages and both genders in the general population and so it cannot be applied to an individual patient [9]. For example a measurement of organ doses and effective dose calculations from a chest radiograph could be the same in a 15 year old and a 35 year old female. Using data published by Wall et al [9] the lifetime risk of cancer incidence for breast tissue in 10-19 year olds of 3.34% per Gray and 30-39 year olds of 1.44% per Gray it can be seen that the risk to female breast in 10-19 year old is higher. This difference in sensitivity due to age and gender is not captured within conventional effective dose calculations.

The tissue weighting factors published by the ICRP are derived using data that is assessed and analysed by The United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR) on cancer risks from follow-up studies of the Japanese atomic bomb survivors [10]. As a result of these on going long term studies the tissue weighting factors have undergone a number of revisions as new data on cancer incidence has been collected [11]. The effect of these revisions can be seen in

Table 3.

Table 3 Comparison of the tissue weighting factors form ICRP publications 26, 60 and 103 [1, 12, 13]

Organs	Tissue Weighting factors		
	ICRP 26 (1977) [12]	ICRP 60 (1990) [13]	ICRP 103 (2007) [1]
Gonads	0.25	0.20	0.08
Red Bone Marrow	0.12	0.12	0.12
Colon	-	0.12	0.12
Lung	0.12	0.12	0.12
Stomach	-	0.12	0.12
Breasts	0.15	0.05	0.12
Bladder	-	0.05	0.04
Liver	-	0.05	0.04
Thyroid	0.03	0.05	0.04
Skin	-	0.01	0.01
Bone Surface	0.03	0.01	0.01
Salivary Glands	-	-	0.01
Brain	-	-	0.01
Remainder	0.03	0.05	0.12
TOTAL	1.00	1.00	1.00

From Table 3 it can be seen that weightings assigned to the gonads have undergone significant changes over the three publications, from 0.25 in ICRP 26 to 0.08 in ICRP 103 that is a reflection of the understanding of heritable risk and the change in breast tissue changed from 0.05 in 1990 to 0.12 in 2007 due to a decision by the ICRP committee to put more emphasis on cancer incidence rather than mortality [4-6]. Brenner, in a number of publications, suggests that these tissue weighting factors represent a subjective balance between the different stochastic endpoints of cancer incidence, cancer mortality, life shortening and hereditary risk. This subjectivity is an example of the “flaws in the science” behind the derivation of these factors [4-6]. However Dietze [11] argues that this revision was in response to the publication of more reliable cancer incidence data published by UNSCEAR [10] rather than a change in the committee’s emphasis. Whatever the reason, it is clear that these revisions do have an impact on effective dose calculations making comparisons to older data difficult.

Lifetime Risk of Cancer Induction

It is with these criticisms in mind that Brenner proposes an alternative to effective dose that can be applied to individual patients - this is referred to by Brenner as “effective risk”. Effective risk considers the life time risk of cancer induction from an absorbed dose of radiation and the equation for this is shown in Figure 3 [4].

Figure 3 Equation for calculating effective risk [2-4]

$$R = \sum_T r_T H_T$$

Where R = Effective risk

r_T = lifetime radiation-attributable tissue-specific cancer risks (per unit equivalent dose to tissue T)

H_T = is the equivalent dose absorbed by tissue (T)

This equation is very similar to that used to calculate effective dose. It is proposed that the tissue weighting factors are replaced with organ-specific radiation-induced cancer risk, such as those published by The Nuclear and Radiation Studies board ([14] or more recently by Wall [9] (a selection of data is shown in Table 4).

Table 4 Life time risks of cancer incidence for males and females by organ and age for a Euro-American population (% per Gy)

Organ (Male)	Age at exposure (y)						
	0-9	10-19	20-29	30-39	40-49	50-59	60-69
Lung	0.65	0.69	0.73	0.78	0.80	0.76	0.61
Stomach	0.93	0.73	0.57	0.43	0.31	0.20	0.12
Colon	1.49	1.22	0.98	0.79	0.60	0.43	0.25
Red Bone Marrow	1.06	1.05	0.77	0.76	0.78	0.65	0.49
Bladder	0.89	0.76	0.65	0.55	0.46	0.35	0.23
Liver	0.56	0.44	0.34	0.26	0.18	0.12	0.07
Thyroid	0.18	0.10	0.05	0.03	0.01	0.01	0.00
Oesophagus	0.12	0.11	0.11	0.11	0.12	0.14	0.15

Organ (Female)	Age at exposure (y)						
	0-9	10-19	20-29	30-39	40-49	50-59	60-69
Breast	4.92	3.34	2.21	1.44	0.84	0.45	0.21
Lung	1.36	1.46	1.58	1.70	1.78	1.72	1.39
Stomach	1.45	1.14	0.88	0.67	0.48	0.33	0.20
Colon	0.73	0.59	0.48	0.38	0.29	0.21	0.14
Red Bone Marrow	0.48	0.48	0.50	0.45	0.77	0.49	0.29
Bladder	0.70	0.61	0.52	0.45	0.39	0.32	0.24
Liver	0.24	0.19	0.15	0.11	0.08	0.06	0.03
Thyroid	0.92	0.52	0.26	0.13	0.06	0.02	0.01
Oesophagus	0.10	0.09	0.10	0.12	0.15	0.21	0.28

The lifetime risk figures (Table 4) are calculated from stronger data as they are based directly on epidemiological studies and not decided by committee [10, 14]. . The organ-specific radiation-induced risk data reflects current knowledge in the biological effects of radiation. The results would be easier to interpret (for example *x* per 1 000 000) for medical imaging professionals and non-medical imaging professionals too. The lifetime risk can be used to provide risk to different genders and age groups

Conveying risk to patients is arguably one of the more challenging aspects the radiography profession has to contend with. To this end Wall suggests using a category based approach to convey the risk from the radiological examination (Table 5) [9].

Table 5 Four broad risk bands for the typical total lifetime cancer risk for patients [9]

Category	Total lifetime cancer risk
Negligible risk	Less than 1 in a million
Minimal Risk	1 in a million To 1 in 100,000
Very Low Risk	1 in 100,000 To 1 in 10,000
Low Risk	1 in 10,000 To 1 in 1,000

Prior to any analysis of risk, dose data has to be collected. Dose data can be measured or estimated. In medical imaging either method can be used to determine dose data in most situations, although there could be occasions where only one method is suitable.

Modelling Dose

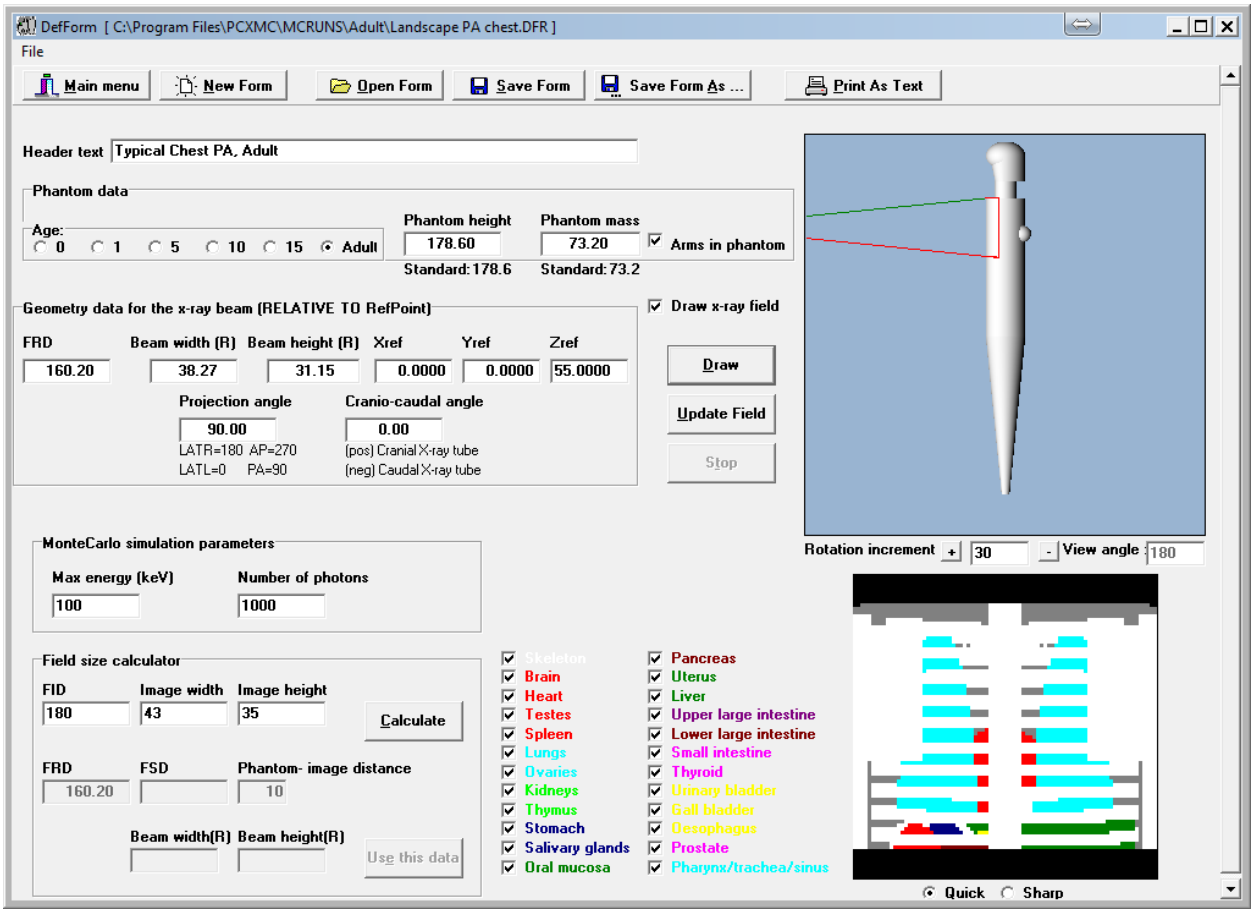
Mathematical modelling of dose using commercially available software is relatively quick and easy. Software is available that allows for organ and effective dose values for conventional radiographic techniques and CT imaging to be estimated. The software employs Monte Carlo modelling which is a mathematical technique that simulates as closely as possible the real interactions suffered by photons.

The process involves the computer simulation of an anthropomorphic phantom being exposed to a large number of photons of varying energies emitted from a point source. The path of each photon is

followed through a sequence of interaction points and subsequent energy losses and outgoing directions (through coherent scattering or Compton scattering). This chain of interactions forms a so-called photon history. At each interaction point the energy deposited to the organ is calculated and used in the dose calculation. A large number of independent random photon histories are generated and estimates of the mean values of the energy depositions in the various organs are used for calculating the dose in these organs Eventually the photon loses sufficient energy to allow photoelectric absorption to occur [15, 16].

PCXMC (STUUK, Helsinki, Finland [16, 17]) is one such programme that allows for organ and effective dose to be estimated in many conventional radiographic techniques. Figure 4 illustrates the positioning of a PA chest with landscape orientation of the image receptor. Other parameters can be manipulated in the software including X-ray anode angle, tube filtration material and thickness to obtain final dose estimates.

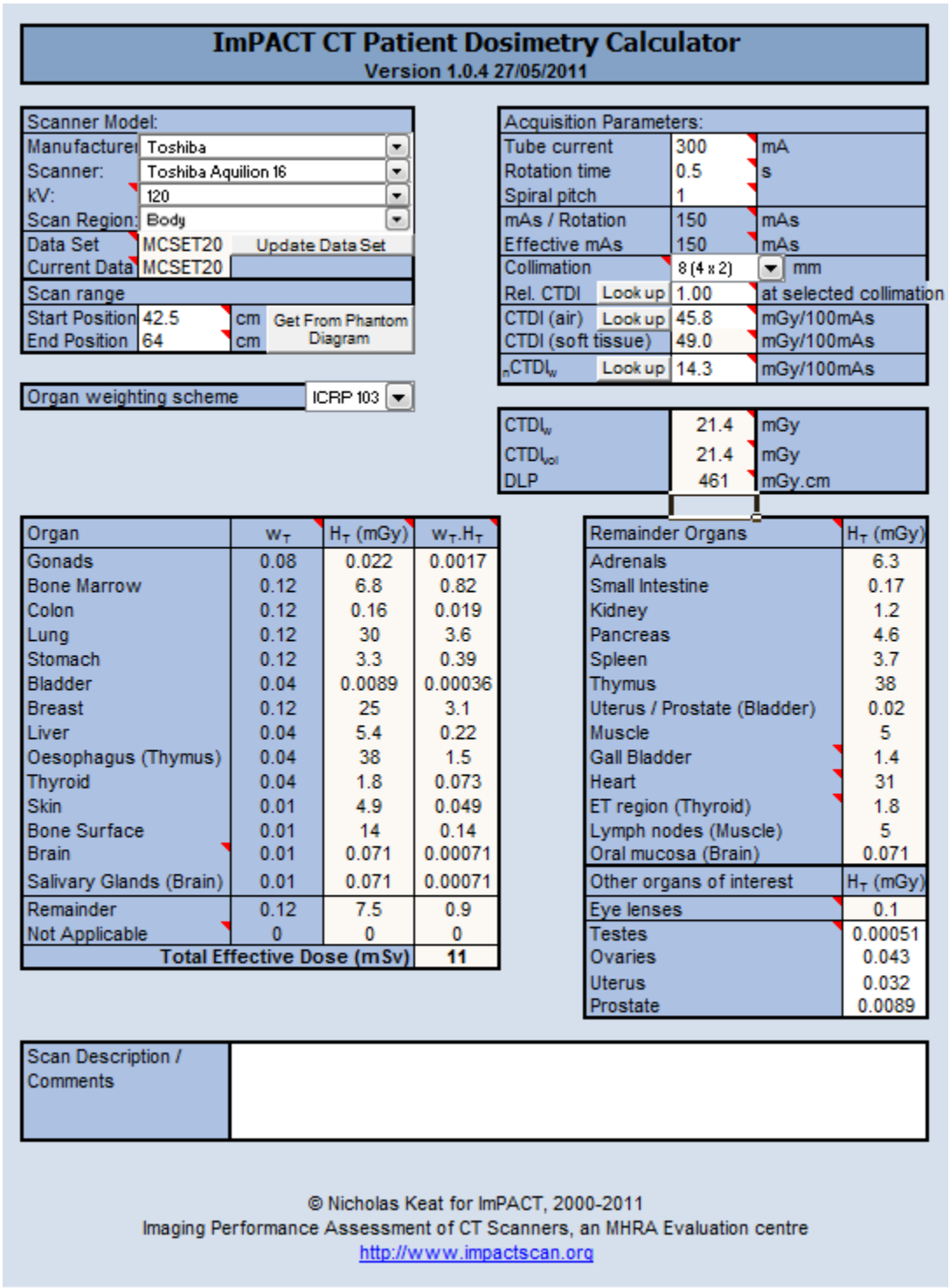
Figure 4 Example of data entry page of PCXMC for the calculation of organ and effective dose from a PA chest radiograph



Dose modelling software is also available for CT dose estimations. For example, ImPact's CT Dosimetry Tool (ImPact, London [18, 19]) software simulation allows for quick and easy calculation of organ and effective dose through the use of Monte Carlo data for normalised organ doses. However, as can be

seen in Figure 5, results are dependent on selecting the imaging parameters and CT model as calculations take into account specific features of each CT unit (e.g. radiation quality and field geometry) [20]. Selection of the correct scanner may not always be possible as new technology and systems are constantly being introduced. These systems are currently not included in dose simulation software meaning that dose simulation has to rely on “best fitting” the attributes of these scanners to those of a similar design. As noted by Groves et al [21], this introduces the potential for significant error in the estimated doses. Automatic mA manipulation by the scanners can also lead to error as the software only allows a single value to be used.

Figure 5 Example of the dose report generated by ImPact CT Dosimetry software



ImPACT CT Patient Dosimetry Calculator

Version 1.0.4 27/05/2011

Scanner Model:			
Manufacturer	Toshiba		
Scanner:	Toshiba Aquilion 16		
kV:	120		
Scan Region:	Body		
Data Set	MCSET20	Update Data Set	
Current Data	MCSET20		
Scan range			
Start Position	42.5	cm	Get From Phantom Diagram
End Position	64	cm	

Organ weighting scheme	ICRP 103

Acquisition Parameters:			
Tube current	300	mA	
Rotation time	0.5	s	
Spiral pitch	1		
mAs / Rotation	150	mAs	
Effective mAs	150	mAs	
Collimation	8 (4 x 2)	mm	
Rel. CTDI	Look up 1.00	at selected collimation	
CTDI (air)	Look up 45.8	mGy/100mAs	
CTDI (soft tissue)	49.0	mGy/100mAs	
$_nCTDI_w$	Look up 14.3	mGy/100mAs	

$CTDI_w$	21.4	mGy
$CTDI_{vol}$	21.4	mGy
DLP	461	mGy.cm

Organ	w_T	H_T (mGy)	$w_T.H_T$
Gonads	0.08	0.022	0.0017
Bone Marrow	0.12	6.8	0.82
Colon	0.12	0.16	0.019
Lung	0.12	30	3.6
Stomach	0.12	3.3	0.39
Bladder	0.04	0.0089	0.00036
Breast	0.12	25	3.1
Liver	0.04	5.4	0.22
Oesophagus (Thymus)	0.04	38	1.5
Thyroid	0.04	1.8	0.073
Skin	0.01	4.9	0.049
Bone Surface	0.01	14	0.14
Brain	0.01	0.071	0.00071
Salivary Glands (Brain)	0.01	0.071	0.00071
Remainder	0.12	7.5	0.9
Not Applicable	0	0	0
Total Effective Dose (mSv)			**11**

Remainder Organs	H_T (mGy)
Adrenals	6.3
Small Intestine	0.17
Kidney	1.2
Pancreas	4.6
Spleen	3.7
Thymus	38
Uterus / Prostate (Bladder)	0.02
Muscle	5
Gall Bladder	1.4
Heart	31
ET region (Thyroid)	1.8
Lymph nodes (Muscle)	5
Oral mucosa (Brain)	0.071
Other organs of interest	**H_T (mGy)**
Eye lenses	0.1
Testes	0.00051
Ovaries	0.043
Uterus	0.032
Prostate	0.0089

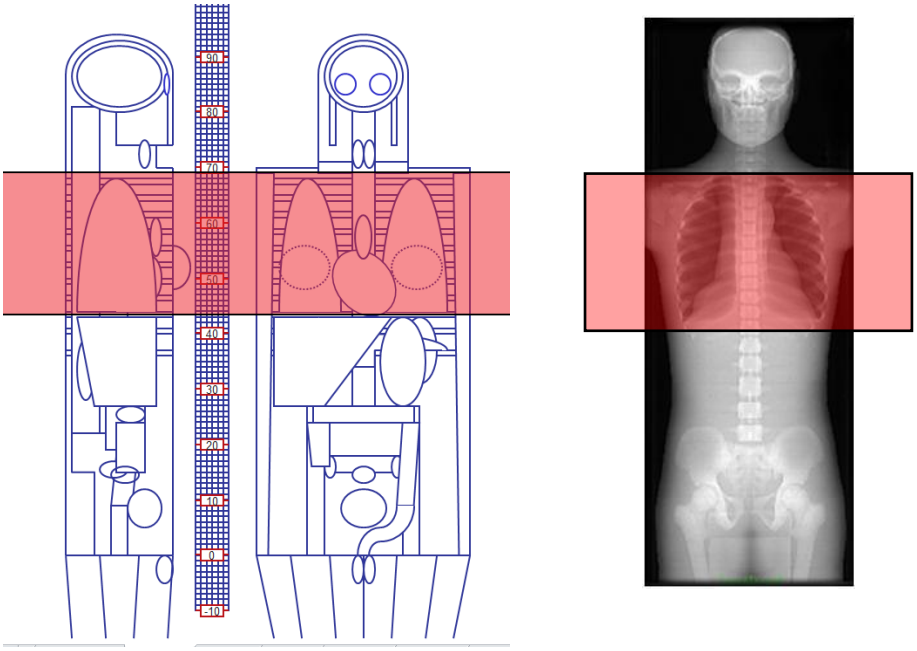
Scan Description / Comments	

Imaging Performance Assessment of CT Scanners, an MHRA Evaluation centre

http://www.impactscan.org

Underestimation of CT doses using computer simulation is frequently reported with magnitudes between 18 and 40% [5, 12, 13]. Reasons for these underestimations have been explained by the differences in the physical dosimetry phantoms and the virtual phantoms used by dose modelling software. Close examination of this highlights the simplified geometric shapes of the organs. Subsequently, as can be seen in Figure 6, in CT examinations of the chest the CT virtual phantom suggests that the liver is not exposed to primary X-ray beam and thus the calculated liver dose would be low. In reality a significant volume of this organ is included in the scan and so will contribute to the effective dose calculations.

Figure 6 Comparison of the virtual phantom in ImPact CT dosimetry software and ATOM dosimetry phantom



However, accuracy of CT dose modelling can be improved by careful selection of the scan range to match the fractions of organs irradiated and to include overbeaming and overranging that is a feature of helical scanning. The use of an average mAs for the scan parameters will further improve the accuracy of the dose calculations [22].

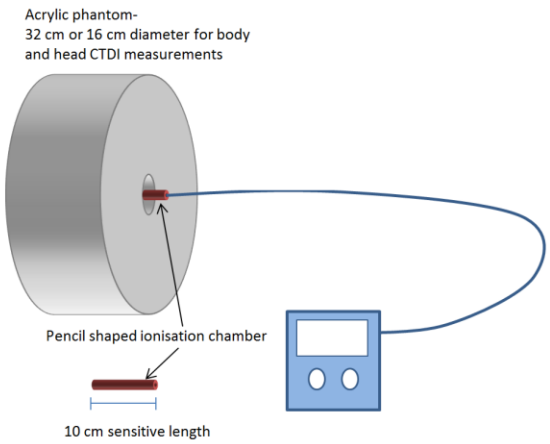
Measuring dose

There are many tools which may be used to measure the dose absorbed during a radiographic procedure that will allow dose to be calculated. The one that most Operators will be familiar with is the Dose Area Product (DAP) meter. DAP combines two quantities- as its name suggests absorbed dose in air and the field size giving the unit Gray centimetre squared $Gycm^2$ (or $cGycm^2$ or $mGycm^2$) (NB not Gray *per* square centimetre). DAP meters are mounted onto the X-ray tube in front of the collimators making readings easy to acquire, however it is important to note that DAP is not patient dose per se

[23]. It is independent of the distance between the source and the patient meaning that if this figure is to be used to estimate patient dose the source to patient distance, the field size and the location of the area exposed are required [24].

CT acquisitions have similar values that are often used as a reference for patient dose; the computed tomography dose index (CTDI) and dose length product (DLP). The CTDI measurement was based on an axial CT scanner and was defined as the dose from the primary beam plus scatter from surrounding slices from a single slice in an acrylic phantom (Figure 7). Phantoms come in two diameters, 16cm and 32cm, to represent the head and body respectively [25].

Figure 7 Typical arrangement of the phantom and pencil beam ionisation chamber used to collect CTDI



Developments in technology and the advent of multislice CT equipment lead to variations in CTDI. $CTDI_{100}$ reflects the dose contribution from a 100mm range (50mm either side of the reference slice). The weighted CTDI ($CTDI_w$) reflects the weighted sum of two thirds the peripheral dose and one third the central dose in a 100mm range. The most commonly quoted CTDI value in modern CT technology is the volume CTDI ($CTDI_{vol}$). This value is obtained by dividing $CTDI_w$ by the beam pitch factor [22, 26]. As before CTDI in any form is not patient dose but a quantification of the radiation output of the CT system so does not take into account differing patient sizes and area of the body that is being imaged [27].

A derivative of CTDI is Dose length product (DLP). This figure takes into account the length of the scan and is calculated by multiplying the $CTDI_{vol}$ by the length of the scan. In a similar way to CTDI and $CTDI_{vol}$, DLP is not patient dose as it does not take into account what part of the body is being exposed, the size of the patient, or the patient's age.

Conversion of DLP to patient dose is possible using a conversion coefficient (*k*) shown in (Table 6). This conversion factor is defined as the effective dose per dose-length product and has the unit mSv / mGy cm. Multiplying the DLP by the relevant conversion factor gives a value for effective dose.

Table 6 Normalised values of effective dose per dose-length product (DLP) over various body regions

Region of the body	Normalised effective dose (E/DLP) (mSv/mGycm
Head	0.0023
Neck	0.0054
Chest	0.017
Abdomen	0.015
Pelvis	0.019

Criticisms of *k* state that the factors are based on old technology and old data; they are based on several scanners that were in use circa 1990 and the tissue weighting factors used in their calculation are from ICRP 60 [13, 26, 28]. There are also a number of assumptions made that would increase the error in the calculated effective dose. For example, the patient is assumed to be standard, and as noted by McCollough et al [27], this standard patient is a little thin by today's standards (nominal body mass of 70 kg). Variation in the way CT scanners report $CTDI_{vol}$ for paediatric patients can make comparison difficult. Some use the 16cm phantom while others use the 32cm phantom. For example Siemens, Philips dose reports are based on a 32cm phantom, Toshiba reports are based on 16 cm phantom and GE reports use the 16cm or 32 cm depending on the scan field of view. $CTDI_{vol}$ can differ by a factor of approximately 2.5 between the two diameter phantoms [27].

True measurement of dose using digital or analogue dosimeters such as metal oxide semi-conductor field effect transistor (MOSFET) or thermoluminescent dosimeters (TLDs) (described later) can be done in a number of ways. In the experimental setting it is possible to measure organ dose by placing dosimeters in a specially designed anthropomorphic phantom. These phantoms are available in a range of patient types; male and female and paediatric, adolescent and adult (Figure 8). They are made up of contiguous slices with different tissues represented by different densities of epoxy resin. The resin has attenuation properties that are equivalent to real tissue. Within the slices are locations for placing dosimeters that will provide data of organ dose (Figure 9). Using these phantoms allows the researcher to carry out experimentation on different techniques, exposure factors or positioning to optimise dose without the involvement of real patients.

Figure 8 The CIRS ATOM dosimetry phantom family models 701-706 (CIRS, Norfolk, Virginia) [29]

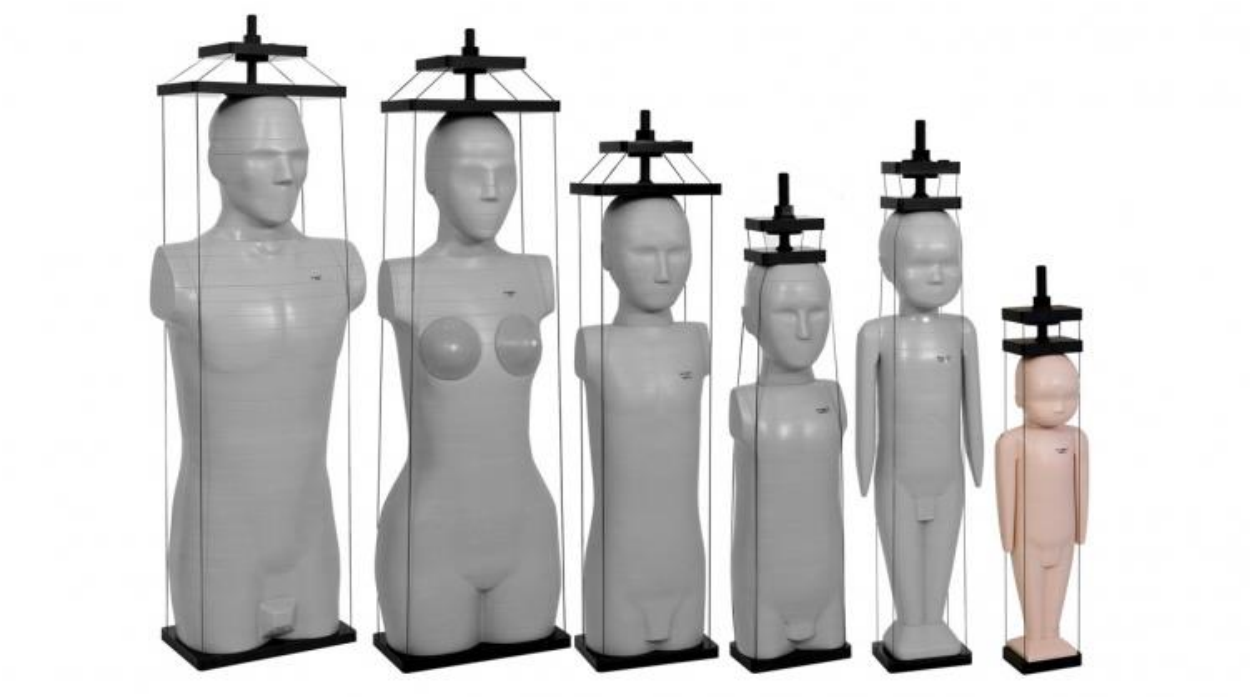
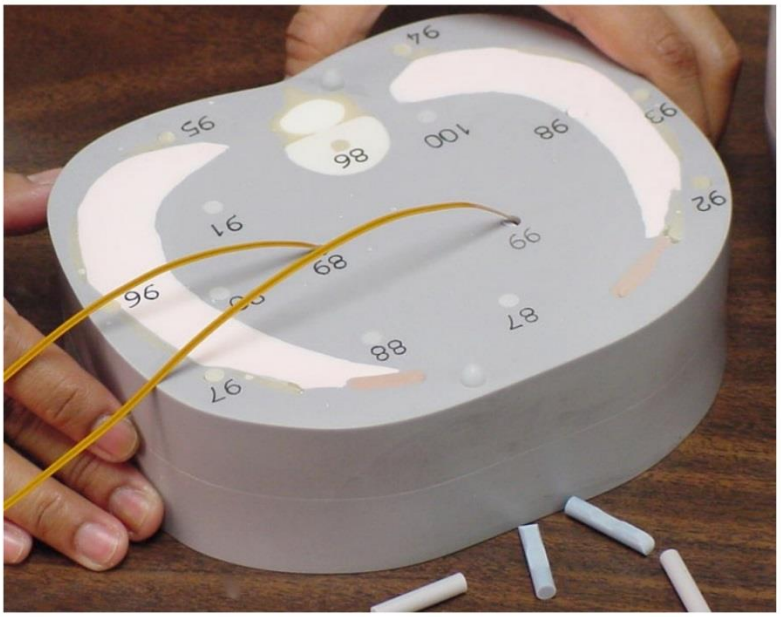


Figure 9 Lower thoracic slices of a paediatric phantom showing different density resin for the lung, soft tissue and bone with locations for dosimeters. These dosimeters can be electronic (shown here by the two wires inserted into the phantom) or analogue such as thermoluminescent dosimeters[29].



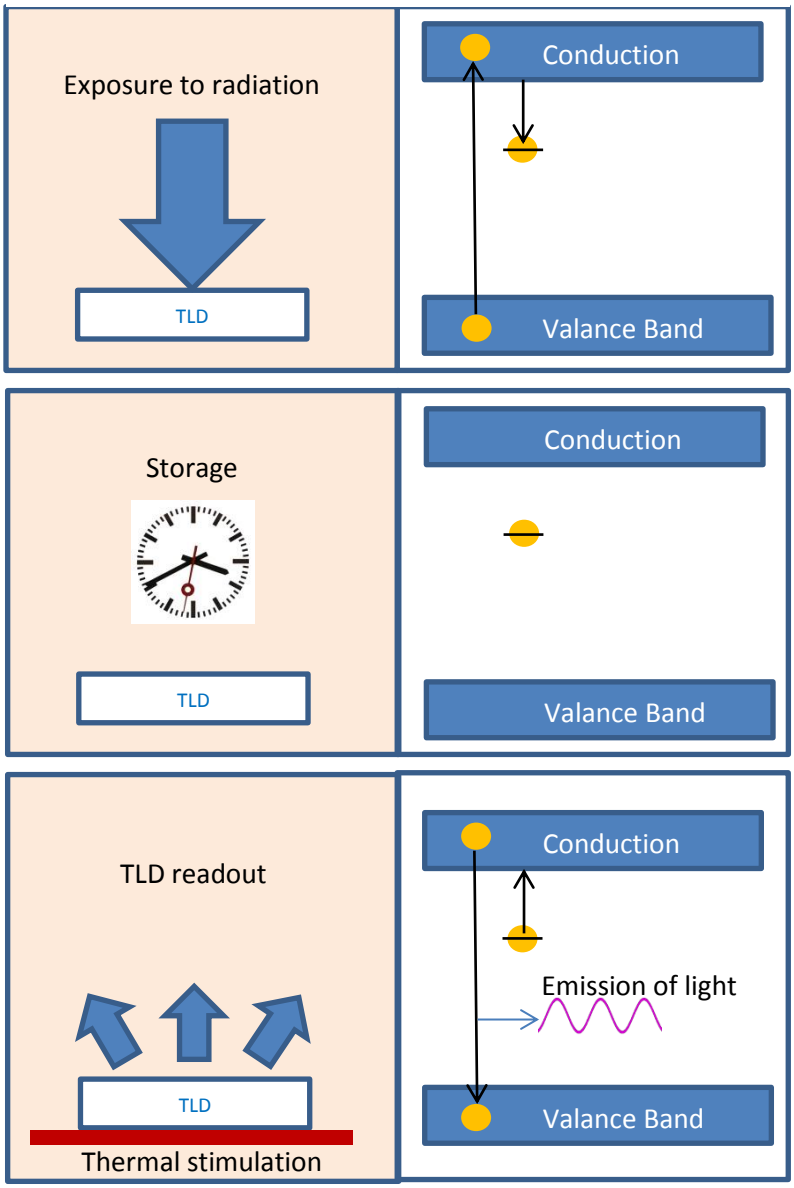
It is obviously impossible to directly measure organ dose in the clinical setting, so the entrance surface dose (ESD) can be used. ESD is defined as the absorbed dose in the skin at a given location on the patient and also includes backscattered radiation from the patient. As a measurement it can be combined with DAP to allow calculations of patient dose to be made.

Dosimeters

ESD and organ dose in the anthropomorphic phantom can be measured using a digital dosimeter or using thermoluminescent dosimeters (TLD). Most medical imaging personnel will be familiar with TLDs in the context of radiation protection as they are frequently used in personal dosimeter badges. TLDs are available in a variety of forms, from powder to square or circular chips, rods, cubes and in a range of materials.

Thermoluminescence (illustrated in Figure 10) uses the atomic model of two energy bands; the valence band and conduction band. Within the valance band electrons are bound to individual atoms as opposed to the conduction band where electrons can move freely within the atomic lattice. Separating these two bands is an area that is referred to as the forbidden gap in which no electron state can exist. The impurities mentioned above create electron traps within this gap. Exposure to ionising radiation excites electrons allowing them to move up to the conduction band leaving holes within the valence band. Electrons can travel amongst the crystal lattice until either the electron can cross back towards the valence band and recombine with a hole or, if near a defect, it can fall into the energy trap. The electron is now prevented from filling a hole within the valance band until it can gain enough energy to once again reach the conduction band before moving back to the valance band. This stimulation is in this context accomplished by introducing heat [30]. The movement of the electron back to the valance band requires the electron to lose energy. This energy is released in the form of visible light and this light is detected by a photomultiplier tube. The charge (measured in Coulombs [C]) generated from this component is measured.

Figure 10 Illustration of the process involved in TLD dosimetry



The choice of material depends on the nature of radiation; in diagnostic and therapeutic energies the chemical composition of the dosimeters is either lithium fluoride with magnesium and titanium impurities added or lithium fluoride with magnesium, copper and phosphorus impurities added [31]. The difference in the materials affects their sensitivity and the measurement range the TLD is capable of. For example TLDs made from Calcium Fluoride Dypromsium are suitable for environmental monitoring and as capable of detecting doses of between 0.1 pGy to 1 Gy [32]. Lithium Fluoride with magnesium and titanium are suitable for medical physics dosimetry applications and operate at doses between 10 pGy to 10 Gy [33].

Conversion from charge to dose involves a calibration process. The TLD or batches of TLDs plus scattering material and a digital dosimeter are exposed to a range of exposures at energy (kV) consistent with the experiment that will be performed. The charge generated from the reading process and the doses recorded by digital dosimeter are used to establish the calibration factor through linear regression. An example of calibration data using is shown Figure 11.

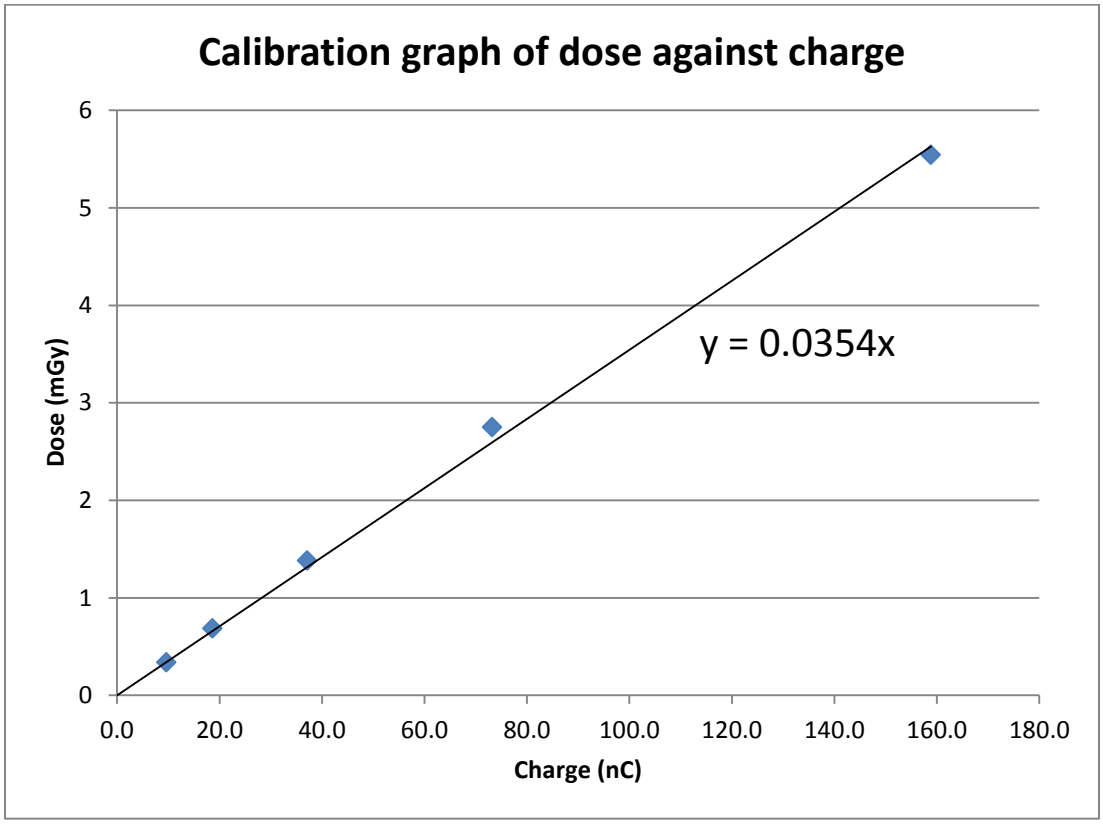
General radiographic equipment can be used in this process although some TLD readers will perform calibration using sealed sources of gamma emitting isotopes such as Strontium-90 or Yttrium-90. Such systems will calibrate each TLD individually rather than in batches increasing the accuracy of the final readings. The response of the TLDs is energy dependent therefore calibration should be performed at the energy that will be used in the research or measurements. If this cannot be done then energy conversion factors can be used but this can introduce error [34].

Figure 11 Example of calibration data for TLDs at 80kV. Plotting data from table (a) results in graph (b) and shows the linear relationship between the charge generated from reading the TLD to dose. The gradient of this line is the calibration factor.

(a)

kV	mAs	Digital Dose reading (mGy)	Background corrected charge (nC)
80	10	0.338	9.7
80	20	0.685	18.6
80	40	1.382	37.1
80	80	2.749	73.2
80	160	5.542	158.9

(b)



One of the disadvantages of the TLD is the time needed to prepare and setup and process them. A typical whole body adult phantom measurement for calculation of effective dose involves the use of 268 individual TLDs. Reading this number using a manual TLD reader equates to approximately 6 hours of work [35]. Research has been undertaken to follow the dental radiography dosimetry process to reduce the number of TLD required for effective dose measurement however, if comparison of organ dose and risk is to be carried out it has been shown that a measurement organ dose is required for all critical organs [35].

An alternative to TLDs is the digital dosimeter. An example of this is the metal oxide semi-conductor field effect transistor (MOSFET) (Best Medical Canada, Ontario, Canada) (Figure 12).

Figure 12 A MOSFET dosimeter with an array of five dosimeters connected to the module [36]

Exposure of the digital dosimeter results in a voltage shift between the components of the dosimeter. This difference is measured and is proportional to the dose absorbed by the detector. However, it is unlikely that the total number of digital dosimeters would be available to allow measurement of all critical organs in one exposure due to expense of the dosimeters. Therefore a number of repeated measurements with the dosimeter relocated between each would be required. As with TLDs, MOSFET dosimeters require calibration and their response is energy dependent meaning separate calibrations are required if a significant difference in beam energies is to be used in any research [37]

Measurements using TLD or digital dosimeters have their advantages and disadvantages relating to preparation time, acquisition time processing following exposure, and cost. These have to be considered when planning research that involves the direct measurement of dose form exposure to ionising radiation [21, 35, 38].

Summary

This article has given insight into terms and concepts associated with dose measurement and modelling, as well as risk estimation. Some limitations and values of dose estimation and measurement methods have been considered. As support for this *special issue* the reader should have gained enough background and insight into Monte Carlo mathematical dose modelling to be able to appreciate some of the empirical articles. Beyond the *special issue* we anticipate that the article could serve as a teaching or CPD aid for personnel working in medical imaging.

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