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H. Järvinen[1], O. Turak[2], M. Ginjaume[3], J. Daures[4], C. Hourdakis[5], P. Ferrari[6], F. Mariotti[6]

[1] Radiation and Nuclear Safety Authority (STUK), Finland

[2]National Research Institute for Radiobiology and Radiohygiene (NRIRR), Hungary

[3]Universitat Politecnica de Catalunya, Institut de Tecniques Energetiques (UPC), Spain

[4] CEA-LIST/LNE-LNHB, France

[5]Greek Atomic Energy Commission (GAEC), Greece

[6]Radiation Protection Institute (ENEA), Italy

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Postfach 1129
D-85758 Neuherberg
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office@eurados.org
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Content:

EXECUTIVE SUMMARY

Management of patient dose in the high dose procedures of interventional radiology (IR) and computed tomography (CT) requires regular and accurate determinations of the practical dose quantities, which are relevant for the basis of actual patient dose determination and setting of the diagnostic reference levels (DRL) or trigger levels for skin dose alert in IR. The accurate measurement of the required quantities is based on accurate calibrations. The complexity of IR procedures and the rapid development of the CT scanners have made an accurate dose measurement a challenging task.

In this report, the recent dosimetric problems as well as the development to cope with these problems have been reviewed. Further, the results of comparisons of KAP (DAP) meter calibrations, both at the level of standard dosimetry laboratories and at hospitals in field conditions, are briefly summarized.

Calibrations of KAP (DAP) systems (meters or computational models) used at hospitals should be carried out regularly to ensure sufficiently accurate basis, within about 10 % (k=2) uncertainty, for the analysis of DRLs, effective doses or trigger levels applied in fluoroscopy. To achieve this, the standard dosimetry laboratories should provide calibrations within about 3 % (k=2) uncertainty for the air kerma; for reference KAP chambers, this should also be the aim while it is recognized to be much more challenging. Calibration laboratories and the users asking for calibrations should consider the clinical conditions where the KAP meter will be used. In the field calibrations, two methods can be used (traditional beam area method and the "tandem" method using a reference DAP meter) but attention should be paid to the particular sources of uncertainty in the method chosen, considering the clinical conditions and the characteristics of the available measuring equipment.

As long as the CT standards retain the present CT dose index concept ($CTDI_{vol}$), the CT equipment manufacturers will follow its implementation in the console readings and dose reporting. Therefore, the rather straightforward, pragmatic approach of the IEC, also proposed as an interim solution by the IAEA, will be the method applied in the basic practice of CT dosimetry for the purposes of patient dose follow-up including the setting of DRLs. Within the pragmatic IEC approach, no change in the basic calibration technique of dosemeters, as recommended in international guidelines (IAEA 2007), is needed. However, the calibration of CT scanners needs to be modified for wide beam conditions, whereby the new IEC approach is recommended. Further studies and experience on its practical implementation in clinical practice is needed. The CT users should be more aware on the correct interpretation on the CT scanner console dose readings and also the importance of the phantom size used in the calibration of the console. $CTDI_{vol}$ and the dose length product (DLP) will continue as the quantities for the setting of DRLs, but further studies should be needed to properly take into consideration also the size of the patient, in particular for paediatric CT. The determination of effective dose for paediatric CT should be carried out with caution, taking into account the limited role of the console dose values. The new dosimetric concepts introduced by the AAPM and the ICRU (Size Specific Dose Estimate, SSDE, equilibrium length, equilibrium dose etc) can become important QC metrics for advanced level users (core groups), but at this stage of CT technology seem to be too complicated for practical use by many of

the users. The practical application of these new concepts requires development of specific software which can automatically calculate the most accurate image-by-image constructed SSDE.

Abstract

This report is the final outcome and summary of the work of the EURADOS WG12, Sub-Group (SG) 3: Technical aspects on DAP calibration and CT calibration.

Management of patient dose in the high dose procedures of interventional radiology (IR) and computed tomography (CT) requires regular and accurate determinations of the practical dose quantities, which are relevant for the basis of actual patient dose determination and setting of the diagnostic reference levels (DRL) or trigger levels for skin dose alert in IR. The accurate measurement of the required quantities is based on accurate calibrations. The complexity of IR procedures and the rapid development of the CT scanners have made an accurate dose measurement a challenging task.

In this report, the recent dosimetric problems as well as the development to cope with these problems are reviewed. Further, the results of comparisons of KAP (DAP) meter calibrations, both at the level of standard dosimetry laboratories and at hospitals in field conditions, are briefly summarized. Based on the review and the comparison of calibrations, a few conclusions and recommendations are given.

1. Purpose and scope

Management of patient dose in the high dose procedures of interventional radiology (IR) and computed tomography (CT) requires regular and accurate determinations of the practical dose quantities, which are relevant for the basis of actual patient dose determination and setting of the diagnostic reference levels (DRL) or trigger levels for skin dose alert in IR. The accurate measurement of the required quantities is based on accurate calibrations. The complexity of IR procedures and the rapid development of the CT scanners have made an accurate dose measurement a challenging task and the users should be aware of the recent dosimetric problems as well as the development to cope with these problems.

The purpose of this report is to review some key aspects of KAP (DAP) meter and CT calibrations and to provide a brief review of the current approaches, or the work going on, in order to improve the accuracy of IR and CT dosimetry. Further, the results of comparisons of KAP meter calibrations, both at the level of standard dosimetry laboratories and in field conditions, are briefly summarized. The results of the review and the calibration comparisons are used to give a set of conclusions and recommendations.

2. KAP (DAP) meter calibrations

2.1 Introduction

Monitoring of patient exposure in diagnostic radiology is necessary for optimization and follow-up of patient doses, in particular for high dose procedures of interventional radiology. Kerma-area product, P_{KA} (the term "KAP" have been used in the past), or dose–area product (DAP), is an important practical quantity which is used to set diagnostic reference levels (DRLs) and as a basis of organ dose and effective dose calculations, via the use of conversion factors. KAP is used both in general radiography and fluoroscopy, to monitor the delivered dose during complex examination requiring several projections of radiography and/or fluoroscopy. It can also be used for setting a warning or trigger level for the possibility of a high skin dose: when this level (KAP value) is exceeded, attention should be paid to the possible skin damage of the patient.

Kerma-area product is measured directly by KAP-meters (DAP-meters) which are thin, wide area, plane-parallel transmission ionization chambers connected to a measuring assembly (electrometer). The chamber can be either removable or fixed in the tube housing. Instead of the actual meter, KAP- (DAP-) values can also be determined computationally based on x-ray tube output and field size settings. Sufficient accuracy for the above KAP-based procedures imposes that the KAP-meters (or computational displays) must be calibrated in an accurate way, traceable to the national standards. This is of particular importance, because the response of a KAP-meter can vary highly (up to 20-30 %) as the function of beam quality (energy) (Toroi et al. 2008, 2011).

The calibration of the KAP meters fixed in the tube housing and the computational KAP displays have to be carried out in situ using clinical beam qualities ("field calibration"). This can be done by using either a traditional method ("beam area method"), where the air kerma and beam area are measured separately and multiplied to obtain the KAP-value, or by using a reference KAP meter calibrated in a laboratory ("tandem" method, Section 2.2) (IAEA 2007, Toroi et al. 2008). In the traditional method, a point like detector is needed for air kerma measurement, and this detector (usually a diagnostic level ionization chamber) has to be calibrated for air kerma in a calibration laboratory.

KAP-meters which are removable can be either calibrated in situ, in the same way as above, or by sending them to a calibration laboratory, where standard beam qualities are usually applied for the calibration. The in situ (field) calibration is generally recommended in order to account for the x-ray equipment-specific characteristics of extra-focal and stray radiation, and it is often also the most convenient method (Shrimpton and Wall 1982, ICRU 2006, Toroi et al 2008). The calibration at the calibration laboratory is needed for a reference KAP meter to be used in the tandem method. The overall accuracy of KAP determinations are affected by the accuracy of the laboratory and field calibrations, and therefore, both class of calibrations have been studied in this EURADOS work and discussed below.

2.2 Basic concepts

The *air kerma–area product*, P_{KA}, is the integral of the air kerma over the area of the X-ray beam in a plane perpendicular to the beam axis (ICRU 2006, IAEA 2007), thus

$$P_{KA} = \int K(x,y)dxdy \qquad (1)$$

The recommended unit is Gy·cm^2 (ICRU 2006), although some other units are often used (Gy mm^2 or µGy m^2) in practice.

When the "*beam area method*" is applied , the above integral is approximated by the product of the air kerma at the centre of the beam and the beam area, assuming sufficient flatness of the beam. Thus air kerma is measured at the centre of the beam using a suitable diagnostic level ionization chamber. The chamber is calibrated at a calibration laboratory using standard RQR calibration beam qualities (IEC 2005), and the calibration factor is chosen on the basis of the measured HVL value of the clinical beam. The beam area (field size) at the calibration distance is determined from a radiographic film or digital image receptor.

In the *"tandem" method of calibration* (Pöyry et al. 2005, IAEA 2007, Toroi et al 2008,), a reference KAP meter is calibrated in a calibration laboratory for the beam incident on the chamber, using standard RQR or other calibration beam qualities. In field calibrations at clinics, the reference and the field chambers are irradiated simultaneously in an x-ray beam passing through both of them. The field KAP chamber is used in the same x-ray unit, position and geometry as in the measurements with patients and the reference KAP chamber is placed at a longer distance. The tandem method avoids the problems of the non-uniformity of the radiation field and the uncertainties of the beam area, provided the air kerma response of the chamber is homogenous and the sensitive area covers the entire beam. The calibration factor of the reference chamber is chosen on the basis of the measured HVL of the clinical beam.

2.3 Accuracy of laboratory calibrations

The accuracy of calibrations in the laboratory, both for ionization chambers for air kerma measurement and KAP plane-parallel ionization chambers for reference KAP measurement, can be assessed and compared based on the estimations of uncertainties of the calibrations, and eventually verified by intercomparison of such calibrations. Within this EURADOS activity, as part of EURAMET project 1177 (EURAMET 2014), a comparison for the calibration of air kerma and KAP ionization chambers was conducted. Besides the calibration laboratories involved in EURADOS WG12/SG3 work, several other European calibration laboratories participated in the EURAMET project (see Section 2.3.2). Standard RQR beam qualities were applied in the EURAMET comparison, but as a supplementary aim within the EURADOS activity, other beam qualities simulating the clinical beam qualities have been studied during the same intercomparison run (Section 2.3.3).

2.3.1 Laboratory level calibration comparison: method

As transfer instruments, two KAP meters were circulated between participating laboratories and the calibration coefficients in terms of P_{KA} and the associate uncertainties were compared. Furthermore, a diagnostic radiology (DR) chamber suitable to measure the air kerma (rate) was circulated and calibrated in terms of air kerma. This made possible to compare separately the differences in the air kerma calibration coefficients and those of the air kerma area product. Conducted as an official EURAMET project, according to the relevant rules, the project enabled participating calibration laboratories to test and verify their calibration methods and capabilities and to support the relevant Calibration and Measurement Capabilities (CMCs) to the International System (SI) at BIPM.

The transfer instruments circulated were as follows:

KAP meters:

- IBA Kerma-X plus (IBA SCANDITRONIX WELLHOFER); Measuring device KermaX-plus DDP TinO, Model 120-205, and KAP Ionization chamber : IBA Model 120-131 TinO.
- Radcal Patient Dose Calibrator PDC (Radcal Corp)

The diagnostic radiology (DR) chamber:

- EXRADIN –Standard Imaging MAGNA A650

The ICRL/GAEC-EIM, Greece acted as the pilot laboratory, which was responsible for the overall coordination and reporting of the comparison. The first calibration was carried out by the pilot laboratory, including constancy checks of the KAP meters and DR chambers. The instruments were then shipped to the participating laboratories in sequence according to the agreed order and time schedule. After every three laboratories, the instruments were returned to the pilot laboratory for an interim re-calibration and constancy checks. The pilot laboratory collected all calibration reports with the associated uncertainty budgets, as well as short descriptions of the calibration procedures. The complete duration of the calibration phase was about 1,5 years. Results from the comparisons were treated according to the CIPM's rules and were strictly confidential during the comparison process.

The KAP meters and the DR chamber were calibrated at the following radiation beam qualities:

- For EURAMET 1177 comparison: Reference beam qualities according to IEC 612674: RQR 3 (50 kV) - RQR 5 (70 kV) - RQR 6 (80 kV) - RQR 8 (100 kV) - RQR 9 (120 kV)
- For EURADOS comparison: Non reference beam qualities to simulate the clinical x-ray beams; on a voluntary basis some laboratories performed the KAP meters and the DR chamber calibration also at the following non-reference beam qualities:

A series: 3.0 mm Al + 0.1 mm Cu at 50 kV, 80 kV, 100 kV and 120 kV

B series: 4.0 mm Al + 0.2 mm Cu at 50 kV, 80 kV, 100 kV and 120 kV

C series: 1.5 mm Al + 0.9 mm Cu at 80 kV, 100 kV and 120 kV

For the comparison of the results, the dependence of the transfer instruments' response on beam energy, the air kerma rate and the irradiation area (for KAP meters) was studied by the pilot

laboratory (in cooperation with some other laboratories), for appropriate correction factors to apply, if necessary.

The uncertainties of the calibrations were evaluated according to the IEC GUM 1995 (IEC 1995) and reported in details at the calibration report of the laboratory. The overall expanded uncertainty were evaluated with coverage factor k=2.

2.3.2 EURAMET 1177 comparison with standard beam qualities: results and conclusions

The KAP meter calibrations (for air kerma area product) were mostly consistent within about 5 % (Degree of Equivalence typically 1-5 %), while in some cases higher differences (up to ~20 %) were observed. Half of the laboratories (11 out of 22) exhibited results that probably were not consistent with the Comparison Reference Value (CRV), since the results within the uncertainties estimated by the laboratory did not include the CRV. The air kerma meter calibrations were mostly consistent within about 1 % (Degree of Equivalence typically 0.1-1.7 %), while in a few cases higher differences (up to ~13 %) were observed. For air kerma calibrations, the results of most of the participants (20 out of 22) were consistent.

The performance of the transfer instruments was tested by the pilot laboratory and a few other laboratories, via repeated checks and measurements. The test results revealed that the characteristics of the transfer instruments were not optimal as transfer: the need for a correction for the dependence of the instrument's response on radiation energy (beam quality correction) was practically negligible, but corrections for the dependence of the instrument's response on the area and air kerma rate were significant in some cases (up to about 60 % in the worst case). Unfortunately special designed, high quality, reference class KAP meters are not commercially available, and the transfer instruments used are widely used for clinical measurements. Due to these limitations, a pragmatic approach was justified: for a meaningful comparison of the calibration capabilities of different laboratories, the effect of the undesirable instrument's characteristics were removed by using appropriate correction factors, and the uncertainty of these correction factors were taken into account in the evaluation of the uncertainty of the comparison values. The details of the results are published by EURAMET (EURAMET, 2015).

2.3.3 EURADOS comparison with non-standard beam qualities: results and conclusions

Clinical diagnostic radiology applications, especially in interventional radiology and pediatrics, often require the use of aluminum and copper (Al + Cu) tube filter combinations as shown in Fig 1. These filter combinations produce radiation qualities that are not covered by the international standards (IEC, ISO). Consequently, the traceability of measurement at these radiation qualities is not well addressed.

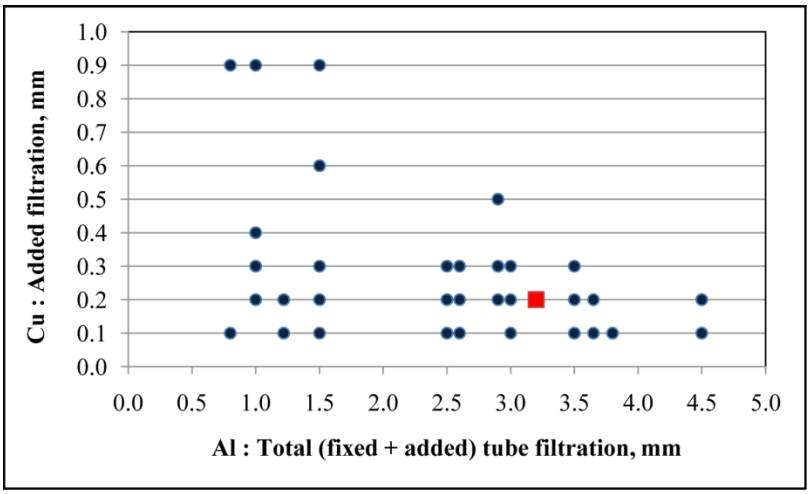


Fig. 1. The results from a survey for the tube filter combinations used at clinics. The (x,y) coordinates of each point correspond to the Al and Cu thickness, respectively. The RQT8 quality (3.2 mm Al + 0.2 mm Cu) is marked in red square.

Depending on the use of the KAP meters, they can measure the incident or the transmitted radiation. The later may require additional provisions for the beam hardening (due to the presence of the KAP in the beam); in this case the traceability of measurements should be considered, as well.

For the above reasons, the KAP meter calibration comparison at a few laboratories was performed at various beam qualities (standard and non standard), as well as for incident and transmitted radiation.

For the calibration with non-standard beam qualities, the results of most of the participants are consistent within the scope of the assigned uncertainties. Compared with the results of calibrations with standard beam qualities, there are no significant differences in the consistency of calibrations between laboratories or in the estimated uncertainties of the calibration factors. However, the uncertainties could be slightly reduced if the standards used in the calibration had a direct traceability to primary standards with non-standard beam qualities. Due to the lack of traceability at the non-standard qualities, the calibration coefficients of the reference chamber at these qualities could be obtained by interpolation or extrapolation of data at RQR, RQT or/and RQA standard qualities, as Fig. 2 indicates. Extrapolation is not recommended, especially when the chamber has significant energy dependence of response.

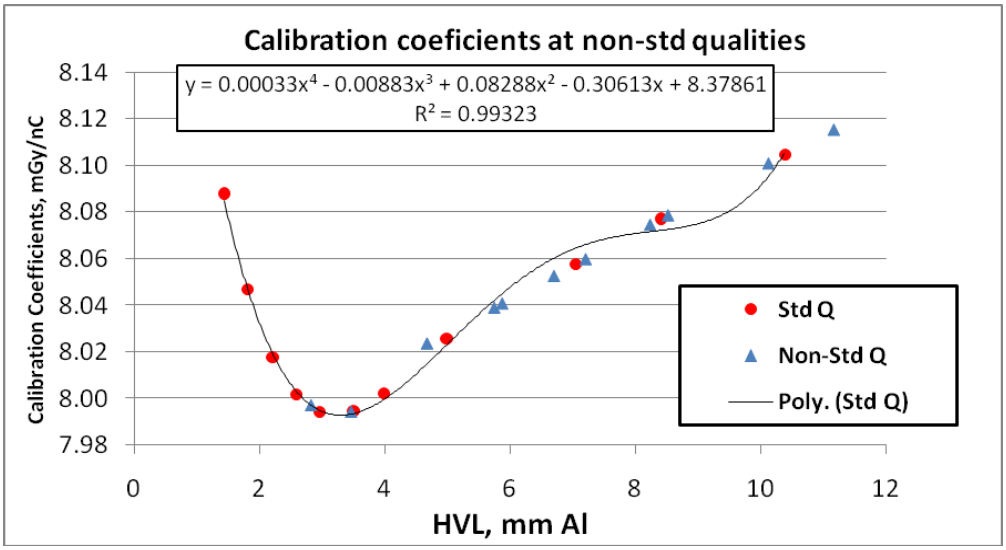


Fig. 2. Example for the determination of the calibration coefficients at non-standard qualities. The red circle points are the calibration coefficient at RQR and RQT standard qualities. The regression of these points is shown in the equation on top. The blue triangle points are the calibration coefficients deduced from regression at the non-standard qualities. The standard uncertainty due to the regression is 2.35 %

Due to the higher range of beam energies towards higher HVL values, and the dependence of the KAP meter's response on beam energy, the calibration factors with non-standard beam qualities can be significantly different from those with standard beam qualities (in particular for Kerma-X meter). Thus, the calibration factors with non-standard beam qualities, simulating the clinical beam qualities, could provide a more accurate KAP value in the clinical measurements. However, due to the generally lower air kerma rates with non- standard beam qualities, results can also be significantly affected by the high air kerma rate dependence of KAP meters, thus jeopardizing the better accuracy achievable on the basis of only beam energy considerations. Further, the comparison of calibrations with non-standard beam qualities becomes very challenging because the problems caused by the non-optimal quality of the transfer instruments are much more prominent.

The results of calibration comparison with the non-standard beam qualities are discussed in more detail in a separate report (to be published).

2.4 Accuracy of field calibrations

Within this EURADOS activity, calibrations of KAP meters or computational KAP displays were carried out at several x-ray units used in clinical practice, using two realistic clinical geometries, over and under couch installations (Järvinen et al. 2012). The two calibration methods, beam area and tandem methods were applied by three calibration teams (Finland, Spain and Hungary).

The reference KAP meter for the tandem method was PDC Version 1.10 (Radcal) in Finland, Diamentor M2 (PTW) in Spain and Diamentor M4-KDK (PTW) in Hungary. In Finland and in Spain,

the reference KAP meters as well as the reference dosimeters for air kerma measurements were calibrated in the secondary standards dosimetry laboratories using RQR beam qualities and in some cases also with extra qualities (simulating heavily filtered clinical qualities). A calibration factor most representative for the clinical beam was used in the measurements (based on HVL, tube voltage and filtration information). In Hungary, a calibration factor provided by the manufacturer (PTW) was used.

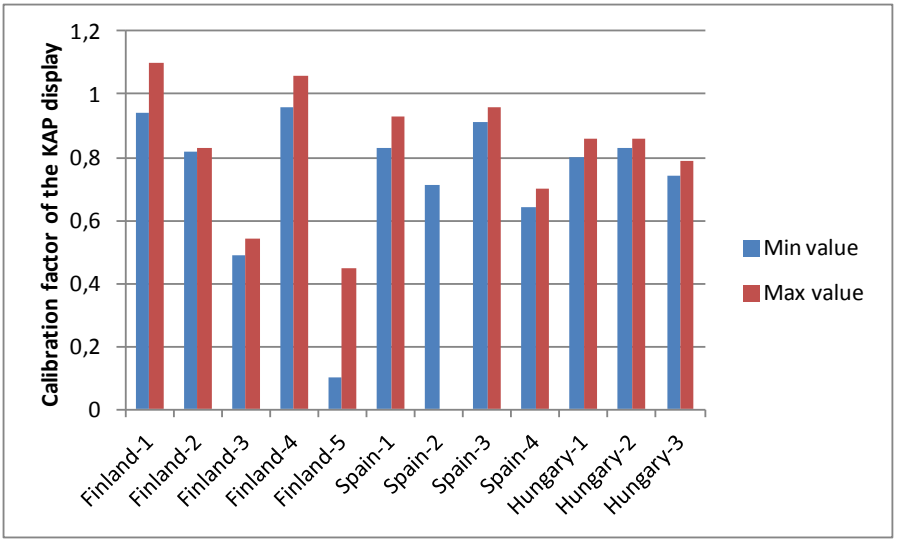


Fig.3. Maximum and minimum calibration factors of the KAP displays for a set of clinical beam qualities and the various x-ray systems used.

The maximum and minimum calibration factors for the KAP displays of the various installations, for a set of typical beam qualities used in clinical practice, are shown in Fig 3 (Järvinen et al. 2012). As can be seen from Fig. 3, the KAP value indicated by the x-ray unit console (based on KAP measurement or computation) can deviate typically 10 % to 40 % from the true value, most often overestimating the real value. This result is consistent with some earlier findings (Vano et al. 2008). Further, the variation of the calibration factor as a function of beam quality, for a given x-ray set-up, was typically within 10-20 %. It is important, therefore, that the local KAP system (for the console reading) has been calibrated before the result is used for the analysis of DRLs, effective doses or trigger levels. The differences between the two calibration methods were generally within about 10 % (Fig.4). For a given x-ray set-up, the difference between the two methods remained rather consistent, generally within a few per cent, with different beam qualities (HVL, kV, filtration).

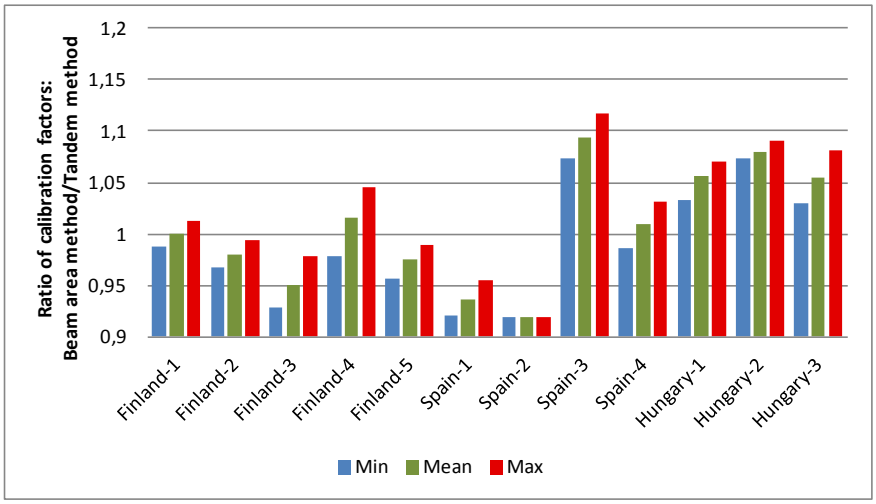


Fig.4. Distribution (min, mean and max) of the ratio of calibration factors: Beam area method/Tandem method.

The typical estimated uncertainties, with careful measurements and taking into account the energy dependence of the reference meters, are about 7 % for both methods when conventional reference meters are used (Toroi et al. 2008, 2011). The uncertainty of the calibration coefficient of the reference meter makes a minor contribution to the overall uncertainty, while other method specific error sources, i.e. difference of calibration and clinical beam qualities for the tandem method, and the measurement of the field area and the effect of field inhomogeneities for the beam area method, make the major contribution to the estimated uncertainty.

The field calibration of KAP meters in interventional units, regardless of the method, turned out to be rather challenging. The calibration may require the operation of the unit in a service mode, with a need of maintenance staff on site, or special arrangements to shield of the image receptor in order to simulate the attenuation of the patient and to enable the system to change the imaging parameters.

2.5 Conclusions and recommendations

The KAP values indicated by the x-ray unit consoles may deviate several tens of per cent from the real value and, therefore, calibration of x-ray unit KAP systems (KAP meter or a computational model) is important for reliable analysis of DRLs, effective doses or trigger levels.

For the calibration of the KAP systems, both the beam area and the tandem methods can be applied with comparable accuracy and convenience in many clinical conditions, with a consistency of methods within about 10 %. The calibration uncertainty of about 7 % or lower (k=2) requires careful laboratory calibration and consideration of the energy dependence of the reference KAP meter in the tandem method, and very accurate determination of the field size in the beam area method. In general, the field calibrations of interventional radiology KAP meters are not

straightforward but may require operation of the unit in a service mode or with special shielding of the image receptor.

Based on the results and discussion in Sections 2.1-2.4, the following conclusions and recommendations can be drawn:

- Calibrations of x-ray unit KAP systems (KAP meter or a computational model) should be carried out regularly to ensure sufficiently accurate basis, within about 10 % (k=2) uncertainty, for the analysis of DRLs, effective doses or trigger levels applied in fluoroscopy.
- Either the beam area or tandem method can be used but attention should be paid to the particular sources of uncertainty in the method chosen, considering the clinical conditions and the characteristics of the available measuring equipment. For the beam area method, the most important factor will be the uncertainty of beam area estimation. For the tandem method, the most important factor is the energy dependence of the response of the reference KAP meter.
- The standard dosimetry laboratories should provide calibrations for at least the air kerma but preferably also for reference KAP chambers within about 3 % (k=2) uncertainty. This will enable sufficiently accurate field calibrations to be conveniently applied. However, due to the non-optimal performance characteristics of the reference KAP chambers, to achieve this level of accuracy can be very challenging.
- The standard dosimetry laboratories should seek to establish additional radiation qualities based on aluminium and copper filter combinations, in order to match the clinical conditions appropriately. Traceability at these new qualities should be achieved and disseminated. This will enable higher accuracy in dosimetry in fluoroscopy and interventional radiology, especially when KAP meters are used.
- The recent X-ray beam specifier based on HVL only, should be reconsidered; tube potential (kV) and X-ray tube filtration should be used, together with the HVL.
- Calibration laboratories and the users asking for calibrations should pay due attention to the clinical conditions (beam quality, dose rate, beam area, radiography/ "continuous" mode/pulsed mode fluoroscopy/cine technique) where the KAP meter will be used.

3. CT Calibrations

3.1 Introduction

There are three main purposes why the radiation output of the CT scanner need to be determined:

- First, as a part of the regular *quality control* procedures, it is important to check, according to an established routine with chosen imaging parameters, that the output of the x-ray tube remains reasonably constant. This gives an indication of the general condition and performance of the equipment: stability of the high voltage, current, x-ray spectrum etc., i.e. the values of the parameters which form the basis for high quality, optimized imaging techniques.
- Second, it is important to follow-up the patient dose levels in order to ensure proper optimization of procedures and the implementation of the good practices. This is generally done by comparing the average patient doses for a group of patients with an established local or national *Diagnostic Reference Levels* (DRL), for given most common and important examinations. This purpose generally imposes that x-ray unit (CT scanner) has an appropriate indicator of the patient dose.
- Third, in order to be able to compare different techniques as for their radiation risk (carcinogenic risk), or the risk of the x-ray imaging with the other risks of life, it is necessary to determine the organ doses (equivalent doses) for the basis of *risk assessment*. The effective dose, calculated from known organ doses with published weighting factors (ICRP 2008), is also often used while it is recognized that it has severe limitations due to differences of patient age and sex distributions from that of general population.

The above purposes call for different type of physical quantities to be determined. For quality control purposes (1), it is only important that the quantity is a reliable indicator of the characteristics, or the changes of the characteristics, of interest for the x-ray system being tested. It should be determined in a highly constant manner while its relation to the actual patient dose is not crucial. $CTDI_{free\text{-}in\text{-}air}$ (Section 3.2.1) is typically applied for this purpose.

For the comparison of patient doses with DRLs (2), quantities that bear a clear relationship to patient dose are desired, while again, in principle, these do not need to be exact indicators of actual doses provided these can be defined and used in a comparable manner. In practice, because patient doses acquired from displays of the x-ray units (or automatically through DICOM header information) are often used also for the doses to be converted to effective doses, it is important that the displayed values have a reasonable relation to the actual patient doses.

For the third purpose (3), the most accurate estimation of the organ doses (and effective doses) requires computational methods which take into account the beam characteristics and spatial distribution in a patient, besides the beam intensity (output). As an approximate method, conversion factors have been determined and published which can be used to convert a measureable (displayed) patient dose (e.g. dose length product, DLP) to effective dose.

A particular aspect in CT is the fact that CT scanners are calibrated based on measurements in standard CT phantoms (Section 3.2.3), so the displayed dose values are only approximating the real

doses to patients of varying sizes. This has an important consequence especially for the determination of patient doses in paediatric CT examinations, where the range for the size of the patient is very high. The recent development has addressed this problem and corrections factors for the size of the patient have been introduced (see Section 3.3).

3.2 Status of dosimetry

3.2.1 Basic dosimetry concepts

For the basic dosimetry concepts, in general, the recommendations of ICRU Report 74, 'Patient Dosimetry for X Rays Used in Medical Imaging' (ICRU 2006) should be adopted, as has been done in the IAEA Report TRS 457 'Dosimetry in Diagnostic Radiology: An International Code of Practice', (IAEA 2007). In both documents, air kerma is utilized as a basis for diagnostic radiology dosimetry.

Table 1 shows the definitions and nomenclature used by the IAEA (consistent with that of the ICRU) for the dosimetry applications for CT in comparison to those used by the IEC. The universal acceptance of IEC nomenclature by CT scanner manufacturers has guaranteed its common usage amongst clinical users of CT dosimetry. The nomenclature of the IEC is used in Section 3.3.3, where the current IEC approach is introduced.

Table 1. Basic dosimetry concepts in CT: Comparison of IAEA (IAEA 2007) and IEC (IEC 2010) dosimetry terminology (adopted from IAEA (2011b).

Quantity	IAEA	IEC
	Measured free-in-air:	
CT air kerma index	$C_{a,100} = \frac{1}{N.T}\int_{-50}^{+50} K(z)\,\mathrm{d}z$	The general equation: $CTDI_{free\ air} = \int_{-L/2}^{+L/2} \frac{D(z)}{N \times T}\,dz$ For beam widths less than 60 mm: $CTDI_{free\ air} = \int_{-50/2}^{+50/2} \frac{D(z)}{N \times T}\,dz$
	Measured in standard phantom:	
Weighted CT air kerma index	$C_{\mathrm{W}} = \frac{1}{3}\left(C_{\mathrm{PMMA,100,c}} + 2\ C_{\mathrm{PMMA,100,p}}\right)$	$CTDI_w = 1/3\ CTDI_{100,c} + 2/3\ CTDI_{100,p}$
Normalized weighted CT air kerma index	${}_nC_{\mathrm{W}}$	${}_nCTDI_w$
Volume CT air kerma index	C_{VOL}	$CTDI_{VOL}$
CT air kerma-length product	$P_{\mathrm{KL,CT}} = \sum_j {}_nC_{\mathrm{VOL}_j} L_j P_{\mathrm{It}_j}$	$DLP = CTDI_{VOL} . L$

Here K(z) and D(z) is the air kerma and dose profile, originating from one axial scan, along a line that is perpendicular to the tomographic plane, dose being expressed as absorbed dose in air. N is the number of active acquisition channels (detector rows), and T is the nominal thickness of each acquisition channel (detector row, or group of detector rows). The CTDI is usually obtained using a single axial scan.

The five CTDI measurements, one in the center and four in the periphery of the CT dose phantom are used to yield one CTDI value, called the weighted CTDI ($CTDI_w$). The weighting factors for deriving $CTDI_w$ are 1/3 for the center CTDI and 2/3 for the averaged peripheral CTDI. With the introduction of helical CT scanners, the volume CTDI ($CTDI_{vol}$), defined as $CTDI_w$ divided by the helical pitch, has been developed to take into account the effect of couch translation (during irradiation) and the associated helical pitch. Conceptually $CTDI_{vol}$ represents the average dose in the central z-axis region of a scanned volume whose length is equivalent to the integration length in the CTDI equation. The region is defined by the couch increment per rotation (or couch incrementation for axial scanning).

The normalized values ($_nC_w$) are the mean values divided by the current-time product.

With the help of Table 1, the equations presented in IEC nomenclature (Section 3.3.3) can be rewritten in conformity with the nomenclature of the IAEA. For example, Eq. (5) can be written as:

$$C_{w,N.T} = C_{w,\mathrm{Re}\,f} \times \left(\frac{C_{a,100,N.T}}{C_{a,100,\mathrm{Re}\,f}} \right) \qquad (2)$$

where $C_{w,N.T}$ is the weighted CT air kerma index for a beam width of N*T mm (if N*T is > 40 mm), $C_{w,Ref}$ is the weighted CT air kerma index for a reference beam width of 20 mm (or closest possible below 20 mm), and similarly $C_{a,100,N.T}$ is the CT air kerma index measured free in air with a 100 mm integration length for a beam width of N*T mm and $C_{a,100,Ref}$ is a similar quantity at the reference beam width.

3.2.2 Dosemeter calibrations

3.2.2.1 General principles

The determination of dosimetric quantities relies on accurate radiation measurements with a dosimeter, consistent with IEC standards (IEC 1997), and with a calibration traceable to a standards laboratory. Typically an ionization chamber detector connected to an electrometer is used for radiation measurement in CT. The measurement uncertainty should be determined by the user through an analysis of measurement errors including the calibration uncertainty (IAEA 2011a).

The output quantity of a pencil chamber as used in CT dosimetry is, for some chambers, the kerma (dose in air) length product (P_{KL}). To be consistent with accepted metrology standards the calibration of such a chamber in a standards laboratory should utilize an aperture of defined length to irradiate the central section of the chamber, as described in an international guidance (IAEA 2007).The calibration coefficient for radiation quality Q is then defined as

$$N_{P_{KL,Q}} = \frac{K\, w\, d_r}{M\, d_a} \qquad (3)$$

where M is the reading of the chamber corrected to reference temperature and pressure; K is the air kerma at the point of test; w is the aperture width; d_r is the distance between the focal spot and the point of test; d_a is the distance between the focal spot and the plane of the aperture. The units of the coefficient are then seen to be kerma length product divided by the measurement units. It should be noted that a significant number of CT dosimetry systems have not been calibrated in terms of kerma length, but instead are calibrated in terms of air kerma (dose). In this case, to obtain air kerma (dose) length from the air kerma measurement, the reading must be multiplied by the length of the chamber used for the measurement.

3.2.2.2 Present calibration practices

Despite the general guidance above, different practices have been applied for the calibrations of CT pencil chambers. In some cases, calibrations have been performed by irradiation of the total chamber length in the beam. Therefore, within this EURADOS activity, a questionnaire on CT chamber calibrations was distributed to several European countries. Furthermore, because many countries had no calibration arrangements for CT chambers but rather relied on the chamber calibrations by the dosemeter manufacturers, the questionnaire was also distributed to all known manufacturers of CT chambers.

The questionnaire was replied by 14 countries but only by one manufacturer. In half (7) of these countries, calibrations of CT chambers had been carried out, but only 2-6 per year. In another half, no calibration arrangements were available and the practice seemed to be to carry out recalibrations (when needed) by sending the dosimeters to the manufacture for recalibration. Three of the countries performed calibrations in accordance with the IAEA Guide (IAEA 2007), others used its national modification or appropriate standards (ISO, IEC). The calibrations were carried out in terms of air kerma-length product (KLP) in 5 countries while 4 countries had used air kerma (K_a). Recalibration was a requirement in 6 of the countries, and the calibration interval varied from 1 to 5 years. As the beam qualities in the calibrations, standard RQA, RQR and RQT qualities and also some own qualities were reported. The reported irradiation field widths (3-30 cm) and heights (5-30 cm) indicate that some of the laboratories calibrate the chambers by irradiating a section of the chamber by a narrow beam while others irradiate the whole chamber in the beam. The check of the uniformity and angular dependence of the chamber response was generally not performed during the routine calibration, only one laboratory reported to carry out the check of uniformity.

In the calibration by the one manufacturer who replied, the calibrations were carried out in terms of KLP, by irradiating the whole chamber in the beam, with the reference chamber simultaneously in the beam. RQA and RQR beam qualities were applied. The traceability of the calibration was to a Primary Standard Dosimetry Laboratory and the typical estimated uncertainty of calibration factor was 2,4 % (k=2).

3.2.3 Calibration of CT scanner dose displays

It is a standard requirement that CT scanners indicate the dose delivered to the patient during the CT scan (IEC 2009). In the modern scanners, usually the $CTDI_{vol}$ and DPL are indicated in the scanner console. The calibration of the console readings are usually carried out by using a pencil ionization chamber of 100 cm length and standard CT phantoms, cylindrical phantoms of 16 cm (head phantom) and 32 cm (body phantom), made of PMMA. The calibration is typically carried out by

measurements of a single axial scan and the calibration factor obtained is subsequently used for all scan configurations.

The body phantom is applied for body scan modes or large SFOW values while the head phantom is applied for head scan modes or small SFOW values. It is important for the user to recognize which one of the two phantoms have been used for the calibration of the different scan modes or scanning protocols; this information should be available and confirmed by the manufacturer. In general, most scanners use the 32 cm phantom on all (adult and pediatric) body protocols, but a few scanner types use the 16 cm data for pediatric body protocols; in some cases pediatric body protocols are calibrated with either of the phantoms depending on the size of the patient. There is a difference in the displayed value by a factor of two depending on its calibration by a head phantom (higher dose) or by a body phantom (lower dose).

3.2.4 Problems with wide beams

The development of CT scanners with different clinical scan configurations has been very fast in the last ten years (Fig. 5). Wider and wider beams of the modern multislice and cone beam scanners have created the problem for the measurement of the CTDI with the 100 mm pencil CT ionization chamber or a problem for the concept in general.

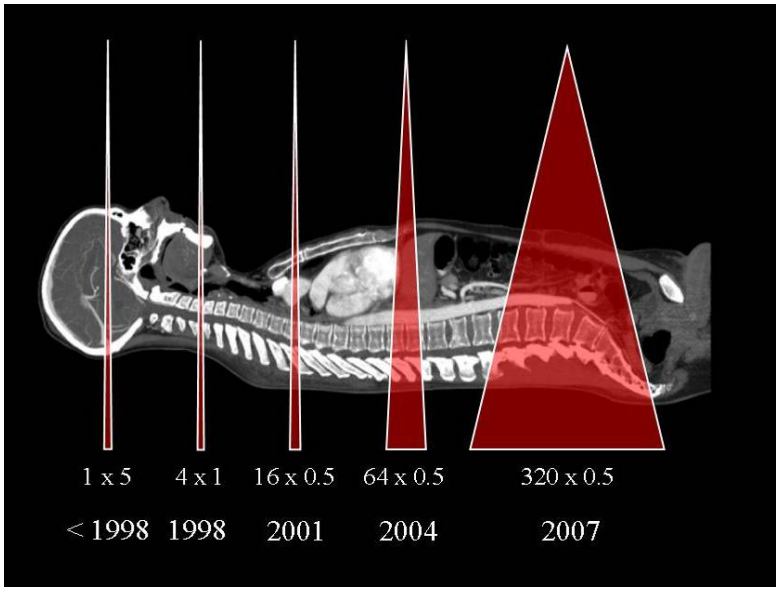


Fig. 5 Progression of typical clinical scan configurations (n*x mm, e.g. 320 * 0,5 mm=16 cm beam width)) shown with the year of the introduction of associated new technology. From IAEA (2011b).

Assessment of CTDI in phantoms has been the subject of much discussion and many revisions since its initial definition. The original definition of CTDI by Shope et al. (1981) was based on integration from -∞ to +∞. Later, the concept was adapted to an integration length corresponding to the actual length of the commonly used CT pencil ionization of 100 mm. Furthermore, since this adaptation, the CTDI has been expressed as absorbed dose in air instead of absorbed dose in polymethylmethacrylate (PMMA).

When CTDI is measured in a CT dose head or body phantom (16 cm and 32 cm in diameter), both the primary beam, and the resulting scattered radiation contribute to the measurement. During the axial scan the primary beam width varies due to divergence at the different peripheral positions within the CT dose phantoms. With a 100 mm pencil chamber, assuming a focus to isocentre distance of 600 mm, the primary beam does not exceed the chamber length during the axial scan at any peripheral position of the CT dose body phantom when the actual beam width does not exceed about 80 mm at the isocentre. Additionally, a field of scattered radiation extends throughout the entire 150 mm long CT dose phantoms, and even further into the surrounding air. Thus a 100 mm CT pencil chamber only includes part of all the scattered radiation. Therefore, for measurements within a CT dose phantom, there is always a discrepancy between the $CTDI_{100}$ and the $CTDI_{\infty}$, where the latter incorporates all the scattered radiation[1]. This ratio is referred to as the CTDI measurement efficiency[2].

The CTDI measurement efficiency depends strongly on the phantom size and on the position of measurement (center or periphery) within the phantom. Fortunately the efficiency remains rather constant for beam widths up to 40 mm. For beams wider than 40 mm the efficiency starts to drop gradually. In conclusion, the current $CTDI_{100}$ metric is not an accurate representation of dose (for scan lengths other than 100mm), for all beam widths.

3.3 Recent progress

3.3.1 AAPM approach

The aim of the AAPM report (AAPM 2010) is finding a way to relate a possible *new dosimetric dose index* to a measurable quantity representative of the air kerma supplied by the new type of CTs during the different *practices*. The main points that pressed toward the following considerations are the availability of multislice detector rows, employed during single acquisition time, the usage of conic beam source and of the possibility, in helical acquisition, of partial superimposition of adjacent slices.

The approach proposed by the AAPM is mathematically committing so only the very general aspects are treated below and a more detailed description, with some added discussion, can be found in the Appendix.

All the theoretical and practical aspects of this method are related to the *Equilibrium Dose*, that is the dose obtained when the scanning length approaches the value of the *Equilibrium Length*. The *Equilibrium Length* concept is derived from the fact that increasing the number of slices of the CT on the left and right of the "central slice", the dose evaluated in the central axis of the "central slice" is influenced by the "queues" of the CT beam impinging on the slices on the right and on the left (that contribute to the radiation scatter cumulated in the "central slice"). It is obvious that, at a

[1]$CTDI_{\infty}$ is the value given by a pencil chamber of sufficient length (∞) to capture the full contribution of scattered radiation generated within a phantom, from a CT axial exposure. Further, $CTDI_{\infty}$ is in principle analogous to the concept of equilibrium dose, (D_{eq}), as described by AAPM report no. 111 (AAPM 2010).

[2]CTDI measurement efficiency is the ratio of the dose integral as limited by the measurement from 100 mm chamber against the total dose integral of a chamber with sufficient length (∞).

certain distance from the "central slice", adding other slices, on the right and on the left, does not augment significantly the measured dose in the "central slice". This produced the concepts of *Equilibrium Length* (that theoretically is ∞) and *Equilibrium Dose*.

There are two more parameters that depend on the employed equipment during the measurement. They are the *geometry efficiency in z direction* (the axial direction of the CT scan), given by the ratio between the product: "number of slices simultaneously acquired in one scan * the slice thickness" and the collimation width and, its reciprocal, the *over-beaming factor*. These quantities are, by definition, related to the collimation width that is an equipment dependent parameter.

Because the *Equilibrium Dose* is influenced by the particular set-up of the scan (given by the combination of kVp, employed filter, number of slice per scan, focal spot, slice thickness, pitch factor, distance between two consecutive slices) it is worth to have an additional parameterization of the *Equilibrium Dose.* The AAPM team demonstrates that is possible to join the *Equilibrium Dose* and *the geometry efficiency in z direction or* the *over-beaming factor* and that reduces the number of combination (and measurement) to be performed in order to characterize the CT equipment. This demonstration allows to employ such parameterization (for which some information from the manufacturer is mandatory) for the practical aspects (measurements).

As a matter of fact the *Equilibrium Dose* can be evaluated, through a small pencil chamber (of 0.6 cm^3 volume), in the middle and in the periphery of a cylindrical PMMA phantom and directly in air. The phantom should be irradiated at the different scanning length in order to find the approximate *Equilibrium Length*. This implies that the conventional PMMA phantom, used for QC in CT, is too short and additional PMMA layers should be added at its extremities (no conclusion on the best phantom to be adopted for the measurements was presented in the AAPM Report (AAPM 2010)).

The pencil chamber should be placed aligned to the "central slice" axis. In the different scanning lengths the chamber, inserted in the phantom, is moved with the patient coach and is irradiated directly by the beam and by the radiation scattered from the rest of the phantom.

All this procedure is thought for quality control and acceptance test to better characterize the quality of the radiation produced by a CT of new generation (helical CT with cone beam) and is not intended for the patient dosimetry. Nevertheless, a so called *Integral Dose*, that is the cumulated energy in the phantom during the CT scan, can be calculated starting from the evaluation of the *Equilibrium Dose* (averaged between the central axis and in the phantom periphery).

3.3.2 ICRU approach

The International Commission on Radiation Units and Measurements (ICRU) has a Report Committee on Image Quality and Patient Dose in CT (Boone 2011). The Committee aims to propose methods that are designed to support and promote Monte Carlo (MC) methods, and to facilitate the integration of MC-based dosimetry into clinical patient CT dosimetry. According to this perspective, MC modeling should be the basis for patient CT dosimetry.

The ICRU committee has considered CT dose profiles and "equilibrium dose" as a function of scan length, consistent with the above considerations by the AAPM (Section 3.3.1). In addition, the ICRU committee has pushed the development of real time (RT) dosimeter probes - capable of 1000 HZ acquisition or better - and been successful in getting several vendors interested in the

development of such probes; some RT probes (scintillation detectors) are already available. The RT probe is of practical importance for the application of the new AAPM/ICRU approach: if such a probe is placed e.g. at the center of the AAPM TG-200 phantom and the phantom scanned from end to end, the dose starts low, gets high, then gets low again as the probe measures scatter, then scatter and primary, then scatter again. Integration of this curve from -A/2 to +A/2 as A is increased gives the "rise to equilibrium curve". These data provide practical methods to correct dosimetry estimates for CT scan length; it provides useful beam characterization and scanner output data, which can then be used as input data for MC modeling and for QA/QC testing for scanner acceptance and periodically.

Besides the scan length consideration, the ICRU approach also proposes a correction to the patient dose due to the size of the patient; this has been studied in collaboration with the AAPM and is equivalent to the data published by the AAPM (AAPM 2011) and discussed in Section 3.3.4.

Furthermore, the ICRU approach also aims to combine dosimetry measurements with image quality assessment. The proposed ICRU phantom is a cylindrical polyethylene phantom of 30 cm length and 30 cm diameter and comprises elements for image quality assessment (contrast, spatial resolution, contrast resolution; MTF, noise, noise power spectra) besides the measurements of CT dose metrics[3].

3.3.3 IEC pragmatic interim approach

The IEC has suggested a practical approach (IEC 2010), which can be an interim pragmatic solution for CT dosimetry. The convenience of this approach is that it essentially retains the long applied and familiar concept of CTDI and also the present practice of its measurements with a 100 cm long pencil ionization chamber, together with the use of the standard CT dosimetry head and body phantoms (cylindrical phantoms of diameters 16 and 32 cm, made of PMMA). The IEC approach has been adopted by the IAEA as an interim solution and detailed guidance for its application has been produced (IAEA 2011b).

3.3.3.1 Basic principle

In the IEC approach, beam widths NxT ≤ 40 mm and NxT > 40 mm are considered separately. Here N is the number of active data channels in a stationary axial scan, and T is the nominal thickness of each data channel.

For beam widths NxT ≤ 40 mm, the conventional definition of $CTDI_{100}$ is applied, i.e.:

$$CTDI_{100,N\times T} = \frac{1}{N\times T}\int_{-50mm}^{+50mm} D(z)dz \qquad (4)$$

Where the subscripts[4], to $CTDI_{100,N\times T}$, indicate an integration length of 100 mm, and a nominal irradiation beam width of N×T.

[3] Since the time of this writing, the ICRU has published a new report (ICRU 2012) where the ICRU approach, referred to in this section, has been introduced and discussed in detail.

[4] Note the subscript $N\times T$ has been added to designate the nominal irradiation beam. In this document this is adopted for both phantom and in air measurements. This is additional to the IEC notation where the

For beam widths NxT > 40mm, it is proposed to measure a reference value for CTDI in the standard CT dose phantoms, for a nominal beam width of about 20 mm. This value is then scaled up by the ratio of free in air measurements of CTDI for the wide beam condition, at different collimation settings, and the reference condition:

$$CTDI_{100,N\times T} = \frac{1}{(N\times T)_{ref}} \times \left(\int_{-50mm}^{+50mm} D_{ref}(z)dz \right) \times \left(\frac{CTDI_{free-in-air,N\times T}}{CTDI_{free-in-air,ref}} \right) \qquad (5)$$

where *ref* refers to the reference beam width condition of 20 mm, or closest value below this.

This approach will flatten the curve of the CTDI measurement efficiency parameter as a function of beam width to infinity, as illustrated in Fig. 6. When using Eq. 4 or 5, values of the $CTDI_{100}$ efficiency parameter can be achieved that are consistent for any beam width (Fig. 6, solid line). Therefore the strength of this approach is that even for wide cone beams, the graph of $CTDI_{100}$ efficiency becomes flat as a function of the beam width. The weakness is that Eq. 4 still underestimates $CTDI_{\infty}$, noting that weighted $CTDI_{100}$ efficiency is still approximately 75% in a CT dose body phantom, and far from the 'ideal' 100%.

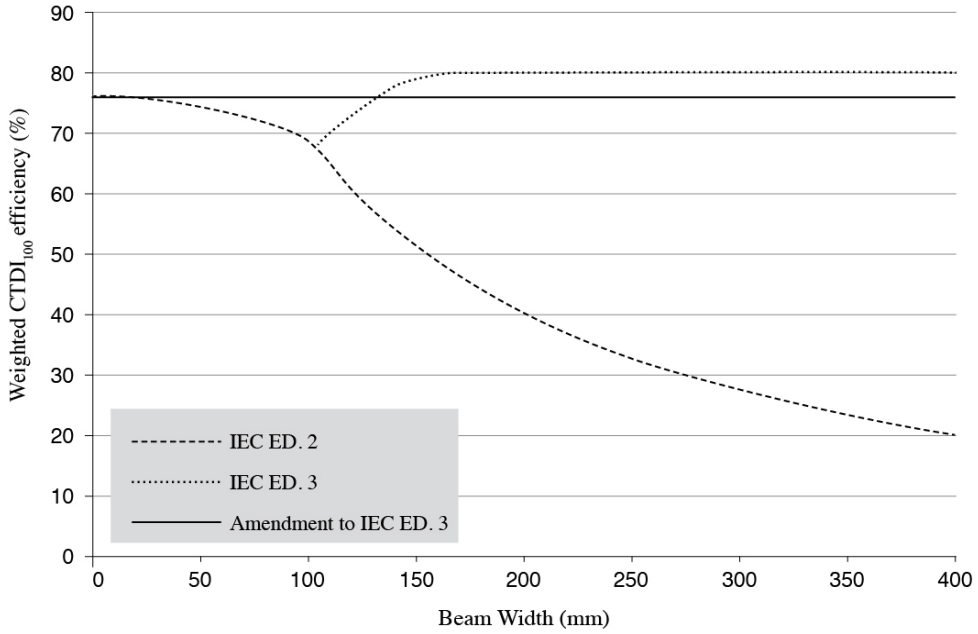


Fig. 6. Illustrative plots of the percentage weighted CTDI measurement efficiency versus beam width for the 150-mm long CT dose body phantom as a function of beam width, for 120 kV and according to three different versions of IEC standards; IEC ED. 3 Amendment 1 refers to Eq. 5 above. The plots are estimated from measurements and from Monte Carlo simulations modelling 100% geometric efficiency in the z direction. The plots are consistent with published data (e.g., Boone 2007, Mori et al. 2005). The three curves correspond to weighted $CTDI_{100}$ as respectively defined in IEC 60601-2-44 Ed. 2, in Ed. 3, and in Amendment 1 of Ed. 3. Graph according to IEC (IEC 2010).

irradiation beam width is only subscripted for the free in air measurements. Note also that the IEC convention of replacing *N x T* by *ref* for the irradiation beam width of the reference beam width has also been adopted.

3.3.3.2 Determination of CTDI

The measurement values obtained from a CT ionization chamber (and the associated dose meter) may be presented as values in units of charge (nC), exposure (Roentgen), air kerma (dose) (mGy), or air kerma (dose) length (mGy.cm). The air kerma (dose) length approach has become more common due to current methods of calibration of the ionization chambers (Section 3.2.2). When the dose meter indicates dose, to obtain the integral in equation (4), the dose value must be multiplied by the length (L_c) of the chamber. When the dose meter indicates air kerma (dose) length, the integral will be directly measured. The measurement values, obtained from the ion chamber and dose meter combination, also need the appropriate correction factors to be applied (calibration factor and the correction for air pressure and temperature).

For nominal beam widths NxT ≤ 40 mm, the $CTDI_{100,\,(NxT)\,\leq\,40\ mm}$ is measured in a PMMA phantom using the traditional method (see IAEA 2007, 2011a). The phantoms to be used are the standard 16 cm and 32 cm diameter, approximately 150 mm long PMMA dosimetry phantoms. The standard approach is for the measurement to be made for a single axial acquisition, with the phantom at the isocentre of the scanner, and the beam centered relative to both the CT dose phantom and the ionization chamber in the z-direction. The measurements are made at the phantom centre position and the phantom periphery positions. They are combined together to give the $CTDI_w$, and when expressed for a helical scanning protocol, the value is multiplied by the pitch factor to give $CTDI_{vol}$ (see Section 3.2.1).

For nominal beam widths NxT > 40 mm, the $CTDI_{100,ref}$ (Eq. 5) is measured in a PMMA phantom using the traditional method for the reference beam width 20 mm (or the nearest beam width available less than 20 mm). To obtain $CTDI_{100,\,(NxT)\,>\,40\ mm}$, CTDI free-in-air need to be determined both for the reference beam width ($CTDI_{free\text{-}in\text{-}air,\ ref}$), using an integration length of 100 mm, and for the desired specific value of N×T ($CTDI_{free\text{-}in\text{-}air,\ NxT}$) with the specified required minimum integration length given in Table 2 (IAEA 2011b). The required minimum integration length is defined to ensure that the integration of the dose profile free-in-air always extends beyond the extent of the beam. The IEC suggests [IEC 2010) that it must be at least 40 mm beyond the nominal beam width (i.e. (NxT)+ 40 mm, or 20 mm at each side of the dose profile), but not less than 100 mm.

For the measurements of the $CTDI_{free\text{-}in\text{-}air,\ NxT}$, the above rule means that for beam widths up to or equal to 60 mm, an integration length of at least 100 mm is required. This length is conveniently provided by the standard 100 mm long ionization chamber, and a single measurement is made with the ion chamber positioned at the central z-axis position of the beam. For beam widths greater than 60 mm, an alternative approach is needed as the required integration distance is longer than the standard chamber distance of 100 mm. The suggested approach is to use the standard 100 mm CT pencil ion chamber, and to step through the beam at regular intervals, usually at a distance equal to the chamber length. A measurement is made at each position. An example of the method is shown in Fig. 7. More precise estimations of $CTDI_{free\text{-}in\text{-}air,\ NxT}$ might perhaps be obtained by employing step sizes much smaller than the chamber length or by uniformly translating the entire chamber through the beam (IEC 2010). More details can be found in the published guidance (IAEA 2011b) and in the IEC document (IEC 2010).

Alternative detectors can also be used, either shorter or longer detectors. With shorter detectors, many more steps are required to acquire the dose along the length of the whole beam. A small (point) detector could be used to record the dose profile from which the integral dose can be

obtained, or by uniformly translating the entire chamber through the beam. Though not widely available, longer ionisation chambers could satisfy the integration length requirements in one measurement.

Table 2. Examples of integration lengths and number of measurements required for $CTDI_{free\text{-}in\text{-}air}$, according to the proposed IEC definition (IEC 2010) with a 100 mm chamber (IAEA 2011b).

Nominal beam width (mm)	Minimum integration length (mm)^	Number of incremented measurements of 100 mm ion chamber	Associated Integration length (mm)
20	100	1	100
40	100	1	100
60	100	1	100
80	120	2	200
160	200	2	200
160	200	3*	300

^ At least, 100 mm or ($N \times T$)+40 mm, whichever is the greater

* The 200 mm integration length is sufficient according to the requirement of IEC, however the 300 mm integration length can also be used since the tabulated length is a minimum requirement.

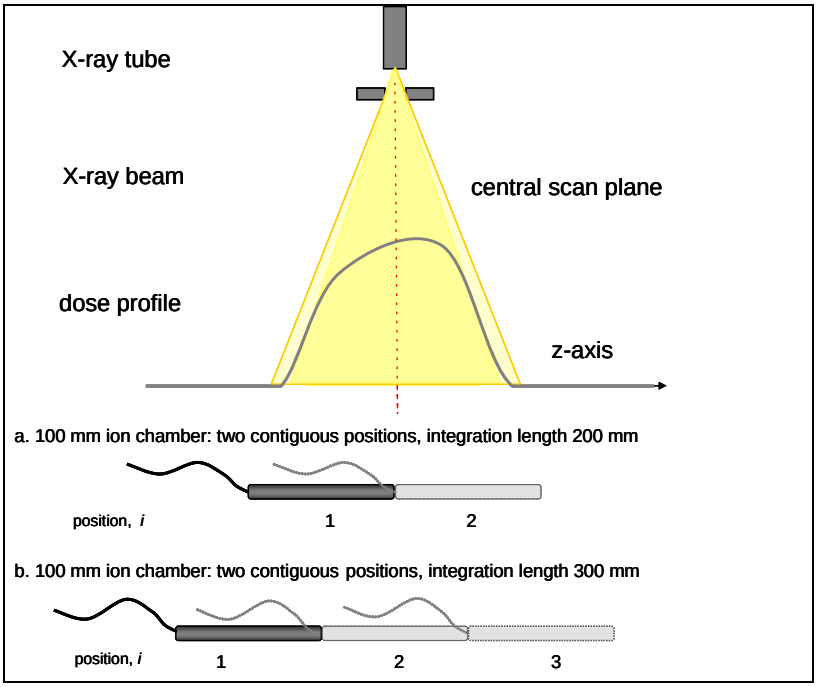


Fig. 7. Diagram demonstrating practical measurements of $CTDI_{free\text{-}in\text{-}air}$ for a beam width of 160 mm with a 100 mm ion chamber, and step increments equal to the ion chamber length. Two integration lengths are shown. The 200 mm integration length is sufficient according to the minimum requirement of IEC, however the 300 mm integration length can also be used (IAEA 2011b).

3.3.4 Patient size correction (AAPM, ICRU)

The present dosimetric concepts of CT, $CTDI_{vol}$ and DLP, are determined for either a 16 cm or 32 cm diameter PMMA cylindrical reference phantom, and therefore, are independent of the patient size. This is of particular importance for pediatric CT because the range of size for pediatric patients is very high. Furthermore, additional confusion is caused by the fact that some manufacturers use the 16 cm diameter phantom and some use the 32 cm diameter phantom as the reference for calculating $CTDI_{vol}$ and DLP (Section 3.2.3). To accurately interpret $CTDI_{vol}$ and DLP for an individual patient, or to compare to other reported values, the phantom diameter used for a specific scanner model and software version must be known. This is a concern, because for smaller pediatric patients, interpreting the displayed $CTDI_{vol}$ (or DLP) as patient dose – without recognizing the distinction between the two phantom sizes – could lead to underestimating patient dose levels by a factor of 2-3 if the 32 phantom was used for reference (AAPM 2011).

The AAPM, in collaboration with the ICRU and the Image Gently campaign of the Alliance for Radiation Safety in Pediatric Imaging, have tried to produce user-friendly computational tools for pediatric radiologists, medical physicists and radiological technologists, to estimate radiation dose during pediatric CT examinations (AAPM 2011). These tools comprise conversion factors that can be applied to the displayed $CTDI_{vol}$ values to allow practitioners to estimate patient dose, and these factors are applicable not only to pediatric CT but to CT examinations of patients of any size.

3.3.4.1 Size related parameters

The anterior posterior (AP) and lateral dimension (LAT), along with effective diameter, is illustrated in Fig. 8 (AAPM 2011). The lateral dimension can be determined from PA or AP CT radiograph, and the AP dimension can be determined by a lateral CT radiograph. In the absence of a CT radiograph, the LAT or AP dimension could be determined using physical patient calibers, which are sometimes available in a radiology facility. The effective diameter corresponds to a circle having an area equal to that of the patient's cross section on the CT image. The effective diameter can be calculated by:
effective diameter = $(AP \times LAT)^{1/2}$

It has also been shown that the sum of AP and LAT dimension (AP+LAT) is linearly related to the effective diameter.

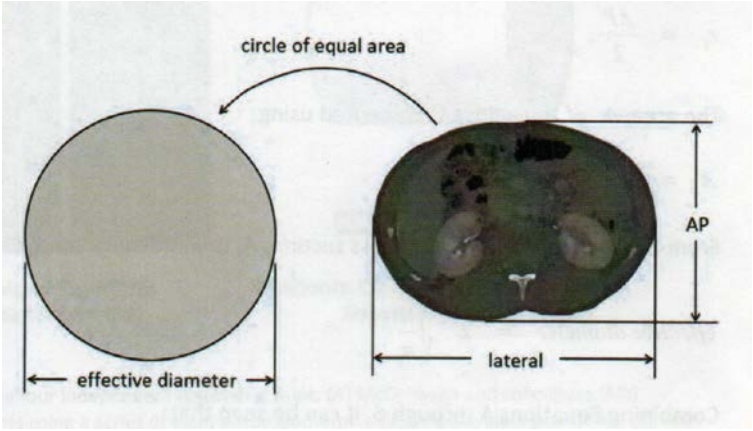


Fig.8. Size related parameters (From AAPM 2011).

3.3.4.2 Size-specific dose estimate (SSDE)

Several research groups using different methods (both experiments and Monte Carlo calculations) have studied the dependence of the mean dose on the central part of the scanned region as a function of patient size. The AAPM Report (AAPM 2011) have combined the results and concluded that, for a given size of the reference phantom for $CTDI_{vol}$ (32 or 16 cm), only one set of size-dependent correction factors (conversion factors) is necessary across tube voltage and scanner manufacturer and model to estimate patient dose from $CTDI_{vol}$. The AAPM has tabulated the conversion factors based on the use of both the 32 cm and 16 cm diameter phantom, as a function of LAT, AP and AP+LAT dimensions as well as of effective diameter. In general, the factors are greater than unity for effective diameters smaller than the diameter of the reference phantom, and lower than unity for larger patients.

When the patient dimension is not known, but the patient age is known, the AAPM provides a relationship of patient effective diameter to age, thus enabling evaluation of the $CTDI_{vol}$ conversion factor also as a function of age. However, this should only be used as an approximation; dose estimates based on patient size are considered more accurate and should be used when size information is available.

For practical nomenclature, the AAPM propose the term "size-specific dose estimate" (SSDE), which should always correspond to tissue doses, not air kerma or other quantities. The specific formula to estimate patient dose for a specific patient size is then given by:

$$SSDE = f_{size}^{32X} \; CTDI_{vol}^{32} \qquad (6)$$

or

$$SSDE = f_{size}^{16X} \; CTDI_{vol}^{16} \qquad (7)$$

where 32 and 16 refer to the size of the scanner calibration phantom, *X* refers to the specific measure of size used (*X*=*S* for AP+LAT, *L* for LAT, *A* for AP and *D* for effective diameter).

3.3.4.3 Limitations on the use of SSDE

The conversion factors given by the AAPM are considered reasonable for use in CT examinations of the torso (chest-abdomen and/or pelvis). However, it is stressed that the use of the conversion factors produce only an estimate of the patient dose at the center of the scanned region (along *z* axis), and it will be most accurate when the patient is centered in the gantry. It is also noted that the average dose in the entire scan is slightly lower than this estimate[5].

The method introduced by the AAPM do not allow the estimation of organ doses *per se*, and in the absence of organ dose data it is not possible to infer effective dose from the methods described. It is also very important to keep in mind that the SSDE is NOT to be used for estimating effective dose based on established conversion factors between effective dose and dose length product (DLP),

[5] Since the time of this writing, the recent ICRU Report (ICRU 2012) has brought about important additional considerations. First, instead of effective diameter, a water-equivalent diameter has been introduced to account for the fact that the effective diameter overestimates the attenuation in chest region where a lot of air is inside. Second, due to the varying anatomy of the imaged reagion and the effect of tube current modulation, instead of the global conversion factor from $CTDI_{vol}$ to SSDE, CT-image-by-image conversions are needed to obtain a more accurate SSDE.

where *DLP = scan length x CTDI$_{vol}$*. This is because such conversion factors are based on computed organ dose data in adults, and it is likely that the organ dose distribution will be different in smaller patients – due to self attenuation issues and differences in how the beam shaping filter influences organ doses as a function of patient diameter.

Finally, the SSDE should not be used to compute a modified dose length product (DLP), i.e. scan length x CTDI vol.

3.4 Impact on calibrations and dosimetry: conclusions and recommendations

As long as the CT standards (IEC 2009, 2010) retain the CTDI$_{vol}$ concept, even though in a proposed modified form, the CT equipment manufacturers will follow its implementation in the console readings and dose reporting. Therefore, the rather straightforward, pragmatic approach of the IEC, also proposed as an interim solution by the IAEA, will be the method applied in the basic practice of CT dosimetry for the purposes of patient dose follow-up including the setting of DRLs.

The following conclusions can be drawn from the considerations in Sections 3.1 -3.3:

- Within the pragmatic IEC approach, no change in the basic calibration technique of dosemeters, as recommended in international guidelines (IAEA 2007), is needed. The CT chambers should be calibrated in accordance with the recommended procedures (IAEA 2007), including partial irradiation of the chamber. As the questionnaire revealed differences of the basic techniques applied, a more rigorous keeping to the recommended technique should be applied in order to increase the accuracy and consistency.
- The calibration of CT scanners needs to be modified for wide beam conditions, whereby the new IEC approach is recommended. Further studies should be carried out and experience collected on
  - The accuracy of the practical method (e.g. the stepping procedure for the determination of CTDI$_{free\text{-}in\text{-}air,\ NxT}$)
  - How this result of calibration is implemented in clinical practice?
- For the calibration of CT scanners, it is important to recognize the calibration phantom diameter for each scan SVOV/protocol.
  - It would be useful if the phantom size to be used in the calibration for given SVOV values could be internationally standardized so that no confusion would be possible between different scanner types.
  - Possibility to introduce only one size of phantom should be further studied (such as proposed by the ICRU)
- The CT users' awareness on the correct interpretation on the console dose readings should be improved by training
  - Recognize difference between "Dose in standard phantom" and the real patient dose
  - Recognize various types of values of the current-time product (mAs)
- CTDI$_{vol}$ and DLP will continue as the quantities for the setting of DRLs. Further studies should be needed on:

- Possibility to apply patient size correction to $CTDI_{vol}$ (SSDE) for setting DRLs for paediatric CT[6]
- Possibility to define a "modified DLP" to account for patient size correction.

› The determination of effective dose for paediatric CT should be carried out with caution, taking into account the limited meaning of the console dose values. Further studies would be needed on

- Possibility to introduce new conversion factors, from corrected $CTDI_{vol}$ (or a " modified DLP") to the effective dose
- Practical patient dose determination based on MC calculations.

› The new concepts of the AAPM can become important QC metrics for advanced level users (core groups), but seem to be too complicated for practical use by many of the users. At the current stage of the technology, they do not provide any added value for setting of DRLs.

[6] Since the time of this writing, recent considerations by the ICRU (ICRU 2012) has indicated that a more accurate CT-image-by-image calculated SSDE is needed to account for the varying anatomy of the imaged region and the effect of the tube current modulation. However, as long as the current CT technology does not provide appropriate calculation softwares for this purpose, the SSDE concept will not provide essential benefits over the $CTDI_{vol}$ for the development of the DRLs.

REFERENCES

AMERICAN ASSOCIATION OF PHYSICISTS IN MEDICINE, Comprehensive Methodology for the Evaluation of Radiation Dose in X-Ray Computed Tomography, AAPM Rep. 111 New York (2010). http://www.aapm.org/pubs/reports/RPT_111.pdf.

AMERICAN ASSOCIATION OF PHYSICISTS IN MEDICINE, Size-Specific Dose Estimates (SSDE) in Pediatric and Adult Body CT Examinations, AAPM Rep. 204, AAPM (2011).

BOONE, J., The Trouble with CTDI100, Medical Physics **34** 4 (2007).

BOONE, J. Private communication 2011.

DIXON, R. L., and BALLARD A.C. (2007). "Experimental validation of a versatile system of CT

dosimetry using a conventional ion chamber: Beyond CTDI100." Med Phys 34(8):3399–3413.

EURAMET (2015). Hourdakis C.J., Csete I., Daures J., Jarvinen H., Mihailescu L.-C., (e), Sochor V., Novak L., Pedersen K.M., Kosunen A., Toroi P., Denoziere M., Büermann L., Megzifene A., Einarsson G., Ferrari P., dePooter J., Bjerke H., Brodecki M., Cardoso J., Bercea S., Ciraj-Bjelac O., Compel J., Glavic-Cindro D., Ginjaume M., Grindborg J.-E. Comparison of air kerma area product and air kerma meter calibrations for X-ray radiation qualities used in diagnostic radiology. Report on the EURAMET 1177 project, identified in the BIPM key comparison database (KCDB) as EURAMET RI(I) – S9.

INTERNATIONAL ATOMIC ENERGY AGENCY (IAEA), Dosimetry in Diagnostic Radiology: An International Code of Practice, Technical Reports Series No. 457, IAEA Vienna (2007).

http://www-pub.iaea.org/MTCD/publications/PDF/TRS457_web.pdf.

INTERNATIONAL ATOMIC ENERGY AGENCY, Implementation of the International Code of Practice on Dosimetry in Diagnostic Radiology (TRS 457): Review of testing results, Human Health Report 3, IAEA Vienna (2011a).

http://www-naweb.iaea.org/nahu/dmrp/publication.asp

INTERNATIONAL ATOMIC ENERGY AGENCY (IAEA), Status of Computed Tomography Dosimetry for Wide Cone Beam Scanners, IAEA Human Health Reports No. 5, IAEA Vienna (2011b).

INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS (ICRU), Patient Dosimetry for X Rays Used in Medical Imaging, ICRU Rep. 74 Bethesda, MD (2006).

INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS (ICRU), Radiation dose and image-quality assessment in computed tomography, ICRU Report No 87, Journal of the ICRU Volume 12 No 1 2012.

INTERNATIONAL COMMISSION ON RADIOLOGICAL PROTECTION, The 2007 Recommendations of the International Commission on Radiological Protection, ICRP Publication 103, Annals of the ICRP (2008).

INTERNATIONAL ELECTROTECHNICAL COMMISSION, Medical Electrical Equipment - Dosimeters with Ionization Chambers and/or Semi-conductor Detectors as used in X-Ray Diagnostic Imaging, IEC-61674, IEC Geneva (1997).

INTERNATIONAL ELECTROTECHNICAL COMMISSION, Medical Electrical Equipment - Part 2-44: Particular Requirements for the Safety of X-Ray Equipment for Computed Tomography, IEC-60601-2-44 - Ed. 2.1 (incl. amendment1), IEC Geneva (2002).

INTERNATIONAL ELECTROTECHNICAL COMMISSION, Medical Diagnostic X-Ray Equipment - Radiation Conditions for Use in the Determination of Characteristics, IEC-61267, IEC Geneva (2005).

INTERNATIONAL ELECTROTECHNICAL COMMISSION, Medical Electrical Equipment - Part 2-44 Edition 3: Particular requirements for basic safety and essential performance of X-ray equipment for computed tomography, IEC-60601-2-44 - Edition 3, IEC Geneva (2009).

INTERNATIONAL ELECTROTECHNICAL COMMISSION, Medical Electrical Equipment - Part 2-44 Edition 3, Amendment 1: Particular requirements for basic safety and essential performance of X-ray equipment for computed tomography, IEC-60601-2-44 - Edition 3, Amendment 1; 62B/804/CD, COMMITTEE DRAFT (CD), IEC Geneva (2010).

INTERNATIONAL ORGANIZATION FOR STANDARDIZATION, INTERNATIONAL ELECTROTECHNICAL COMMISSION, (2008). ISO/IEC Guide 98-3:2008. Uncertainty of measurement –Part 3: Guide to the expression of uncertainty in measurement (GUM:1995).

JÄRVINEN H., TOROI P., PARVIAINEN T., TURÁK O., AND GINJAUME M. Accuracy of KAP displays and the feasibility of two methods for their calibrations. Paper presented at Int. Conf. on Rad. Prot. in Medicine, Bonn, Germany, 3-7 December 2012.

MORI S., ENDO M., NISHIZAWA K., TSUNOO T., AOYAMA T., FUJIWARA H. AND MURASE K. (2005). "Enlarged longitudinal dose profiles in cone-beam CT and the need for modified dosimetry." Med Phys 32(4):1061–1069.

NATIONAL RESEARCH COUNCIL OF THE NATIONAL ACADEMIES, Health risks from exposure to low levels of ionizing radiation; BEIR VII phase 2, Committee to Assess Health Risks from Exposure to Low Levels of Ionizing Radiation, National Academies Press, Washington, DC (2006).

http://www.nap.edu/openbook.php?isbn=030909156X.

PÖYRY P., KOMPPA T. AND KOSUNEN A. Calibration of dose-area product meters for diagnostic x-ray beams, Proc. XXXIX Ann. Conf. Finnish Phys. Soc. , Report Series in Helsinki University of Technology Laboratory (Espoo, Finland) (2005), p. 247.

SHOPE, T.B., GAGNE, R.M., JOHNSON, G.C., A method for describing the doses delivered by transmission x-ray computed tomography, Medical Physics **8** (1981), 488-495.

SHRIMPTON P C and WALL B J 1982 An evaluation of the Diamentor transmission ionisation chamber in indicating exposure–area product (R cm2) during diagnostic radiological examinations. Phys. Med. Biol. 29, 871–8.

TOROI P., KOMPPA T. AND KOSUNEN A. A tandem calibration method for kerma-area product meters. Phys. Med. Biol. 53 (2008), 4941.

TOROI P., HUSSO M. AND KOSUNEN A. Calibration of kerma-area product meters with a patient dose calibrator. In: Standards, Applications and Quality Assurance in Medical Radiation Dosimetry (IDOS). Proceedings of an International Symposium. Vienna, 9-12 November 2010, STI/PUB/1514, Volume 1. IAEA 2011.

VANO E, JÄRVINEN H, KOSUNEN A, BLY R, MALONE J, DOWLING A, LARKIN A, PADOVANI R, BOSMANS H, DRAGUSIN O, JASCHKE W, TORBICA P, BACK C, SCHREINER A, BOKOU C, KOTTOU S, TSAPAKI V, JANKOWSKI J, PAPIERZ S, DOMIENIK J, WERDUCH A, NIKODEMOVA D, SALAT D, KEPLER K, BOR MD, VASSILEVA J, BORISOVA R, PELLET S, CORBETT RH. Patient dose in interventional radiology: A European survey. Radiation Protection Dosimetry 2008; 129 (1-3): 39-45.

Appendix: The AAPM approach (AAPM 2010)

The Dose Profile and the Cumulative Dose

The main characterizing parameter chosen by AAPM is the *Dose profile free in air* **f(z).** The **f(z)** in Fig. 1 depends on the *collimation width* **a** (in Fig. 1 is 11.4 mm).

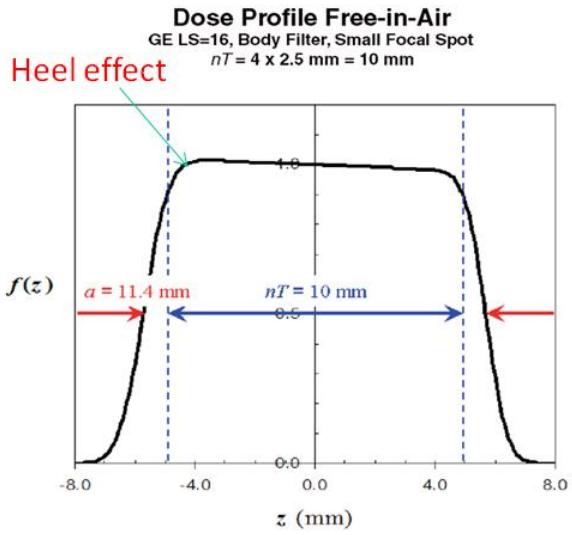


Figure 1

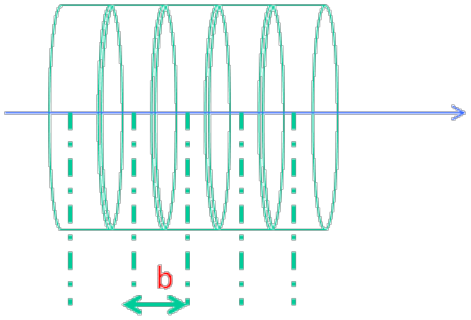


Figure 2

n = number of section acquired in a single scan;

T = thickness of the tomographic section

N = number of rotations;

b = distance between two consecutive slice midpoints;

L=N×b = scanning length

The *Cumulative Dose* $D_L(z)$ is obtained shifting the **f(z)** along the scanning length and superimposing the contribution of the **f(z)** in each point of the scanning (Fig. 3a and 3b). As can be seen from the figures the main effect that increases the $D_L(z)$ is the presence of the queues in the **f(z)** itself and increasing the scanning length, that is adding other **f(z)**s to the sum, the values taken by $D_L(z)$ are consequently increased (Fig. 4).

A valuable index of the behavior of the $D_L(z)$ is its value calculated in **z=0**.

It is easy to understand that this value increases asymptotically to a value called ***equilibrium dose*** and faster than any other point of the scanning length in which the value of $D_L(z)$ is calculated.

Mathematically one can express the value of $D_L(z)$ at $z=0$, $D_L(z=0)$ as:

$$D_L(z=0) = \frac{1}{b}\int_{-\frac{L}{2}}^{\frac{L}{2}} f(z')dz'$$

Where **b** is the distance between two consecutive slices and **L** is the scanning length.

We could now define what happens to $D_L(z=0)$ when the scanning length increases to infinite introducing the concept of *equilibrium dose*: D_{eq} This quantity should be somehow bounded by another quantity: the *collimation width* **a** that is:

$$\lim D_L(z=0) \xrightarrow[L\to\infty]{} D_{eq} = \frac{1}{b}\int_{-\infty}^{\infty} f(z')dz' \propto \frac{a}{b}$$

Introducing the *equilibrium scanning length* L_{eq} that is the finite length at which D_{Leq} is almost indistinguishable from D_{eq} one obtains:

$$D_{L\,eq}(z=0) \approx D_{eq} = \frac{1}{b}\int_{-\infty}^{\infty} f(z')dz' \propto \frac{a}{b}$$

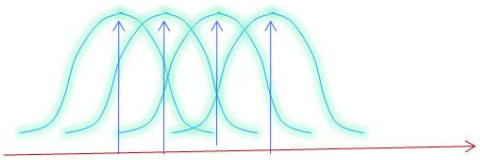


Figure 3a

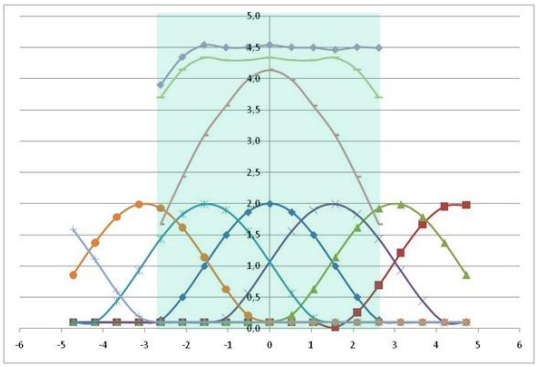


Figure 3b

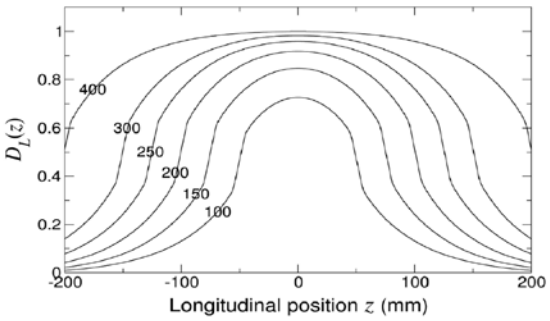


Figure 4

Considering the helical behavior of new CTs it is useful to assume a *generalized pitch factor* **p** defined as:

$$p = \frac{b}{nT}$$

Using **p** one can express the cumulative dose calculated at the equilibrium length D_{Leq} as:

$$D_{Leq}(z=0) \approx D_{eq} = \frac{1}{b}\int_{-\infty}^{\infty} f(z')dz' \propto \frac{a}{b} = \frac{a}{pnT}$$

Determining a and D_{eq} and L_{eq}

If one assumes the introduced metric as the most suitable for the CT, the problem is to determine these new parameters.

The *collimation width* **a** should be obtained from manufacturer through one of the following indexes[7]:

$$\frac{nT}{a} \approx geometric\ efficiency\ in\ the\ z\text{-}direction$$

$$\frac{a}{(nT)} \approx over-beaming\ \ factor.$$

A procedure to determine D_{eq} is not proposed in the document, but the methodology followed by Dixon and Ballard (2007), based on the concept of *approach to equilibrium function*, **h(L)** is suggested.

One can consider the existence of a function **h(L)** for which:

$$D_L(z=0) = h(L).D_{eq} \qquad h(L) \xrightarrow[L=Leq]{} 1$$

Dixon and Ballard established that **h(L)** can be written as: $h(L) \approx 1 - \alpha \exp\left(-4L/L_{eq}\right)$

From this function follows: $D_L(z) = h(L).D_{eq} \approx D_{eq} - D_{eq}\alpha \exp\left(-4L/L_{eq}\right)$

D_L should be measured at various scanning length **L**.

[7] However is important to underline that different combinations of **n** and **T** produce different values of **a** because the penumbra effect, that means that 16x0.625mm; 8x1.25mm,4x2.5mm produce different dose profile.

Then the couples of data (D_L, L) should be plotted trying a fitting with the above function.

From that fitting the corresponding D_{eq} and L_{eq} should be derived.

In Fig. 5 an example of the fitting of the measured data of the above function is presented. All is renormalized by the calculated D_{eq} at peripheral axis; however it is easy to see how the L is 70 mm for the peripheral axis and 112 for the central axis. Indeed the D_{eq} for the central axis is about 0.7 of that calculated for the peripheral axis.

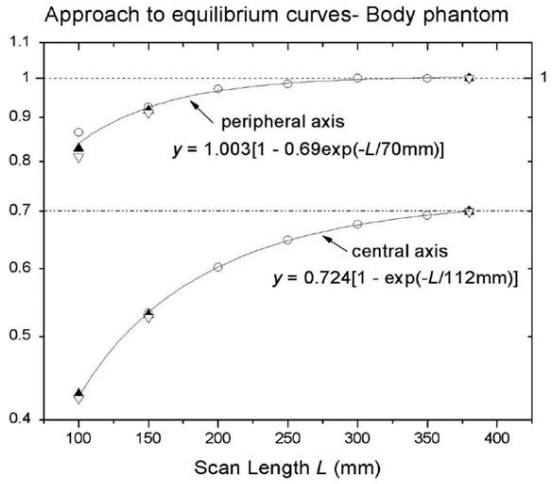


Figure 5

Equilibrium dose pitch product and equilibrium dose constant

It will be shown that the acceptance and periodical QC test procedures require different steps in order to test all possible configuration that could influence the radiation supplied by the CT. For that reason two parameterizations are provided that can reduce the required number of measurements.

The ***equilibrium dose-pitch product*** makes possible to be independent from the pitch chosen, i.e. the pitch employed during the QC test can be different from those used during the clinical application. Because the fitting of D_L varies with different value of p , the product (for $L=L_{eq}$) D_L by p is a constant. As can be seen, this product, called the equilibrium dose-pitch product, is indeed proportional to the over-beaming factor.

$$\hat{D}_{eq} \equiv p.D_{eq}(a,p) \propto p.\frac{a}{pnT} = \frac{a}{nT} \xrightarrow[L=Leq]{} \hat{D}_{L=Leq} = p.D_L(z=0, L=L_{eq}, a, p)$$

The other possible solution is the ***equilibrium dose constant*** that is a value characterizing the tested equipment and can be expressed as:

$$\left(\frac{b}{a}\right).D_{eq} = \frac{p.nT}{a} D_{eq} \equiv \frac{nT}{a} \hat{D}_{eq}$$

The integral dose

The AAPM report is mainly addressed to the acceptance and QC procedures, nevertheless a dosimetric concept linked to the D_{eq} and related to the dose supplied to the patient is given: the ***integral dose***[8], E_{TOT}.

E_{TOT} is defined as the total energy delivered to a CT phantom during the acquisition.

f(z,r) is the single rotation axial dose profile along a given **z** axis, located at distance **r** from the central axis. Then if **N** is the number of rotations and ρ the density one can calculate the total energy deposited in the considered volume as:

$$E_{tot} = N\rho \int_{volume} Dose_{vol} dvol = N\rho \int_{-\infty}^{\infty} \left(\int_{0}^{R} f(z,r) 2\pi r dr \right) dz = bN\rho \int_{0}^{R} \left(\frac{1}{b} \int_{-\infty}^{\infty} f(z,r) dz \right) 2\pi r dr$$

In the last brackets it is easy to recognize the definition of D_{eq}:

$$E_{tot} = bN\rho \int_{0}^{R} D_{eq}(r) 2\pi r dr$$

Now substituting **L=Nxb** and introducing the *planar average* of D_{eq}:

$$\overline{D_{eq}} = \frac{1}{\pi R^2} \int_{0}^{R} D_{eq}(r) 2\pi r dr \quad (9)$$

[8]Here AAPM was not very precise, because the quantity is called *dose* but is an energy.

one finally obtains the *integral dose*:

$$E_{tot} = bN\rho\int_0^R D_{eq}(r)2\pi r dr = L\rho\pi R^2 \overline{D_{eq}}$$

Practical aspects

The AAPM Report explains how to operate in practical QC test following a "guideline" to be repeated for each relevant set (focal spot, bow-tie filter, kVp, **n, T, a**).

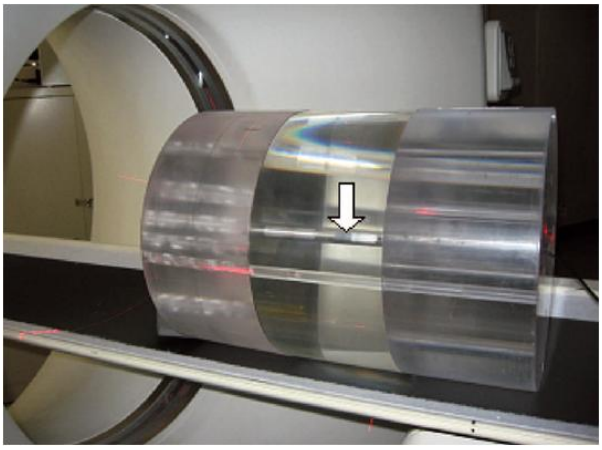
First of all *"start by designating a "reference" set of operating conditions{...}ref to be associated with doses denoted $D_{eq,ref}$ and f_{ref} (0), with parameters focal spot size$_{ref}$, bow-tie filter$_{ref}$, n_{ref}, T_{ref}, a_{ref}, b_{ref}, p_{ref}, etc., and with an approach-to-equilibrium function denoted $h(L)_{ref}$".*

One could use nT/a or a/nT , supplied by the manufacturer, to get a_{ref} then employing, for example, a possible selection of the other parameters such as *Large focal spot*, *Body bow-tie filter*, n_{ref}= 4 slices, T_{ref} = 5 mm .

To calculate $\mathbf{D_{eq}}$ operatively one has to:

- Place the thimble ionization chamber in the phantom central hole. Center the charge-collection volume of the chamber at the phantom central plane, which will correspond to the midpoint (z = 0) of the scanning range (–L/2, L/2) for all measurements. Align the phantom central axis with the scanner axis of rotation. The 0.6-cm^3 ionization chamber (indicated by the arrow, 20-35 mm length) is shown with the middle of its charge-collection volume positioned at the center of the phantom assembly and aligned along the scanner axis of rotation (z-axis direction).

[9] Passages are not described in the text, anyway a very simple formula is introduced for a "classical" cylindrical CT phantom for calculating the *planar average* of $\mathbf{D_{eq}}$ as: $\overline{D_{eq}} = \frac{1}{2}D_{eq}^{\ center} + \frac{1}{2}D_{eq}^{\ perifery}$ About the weighting factors appearing in the formula, D.M. Bakalyar describes in Med. Phys 33(6) why the ½ weighting factors are preferred instead of the 1/3 2/3 employed in $CTDI_{VOL}$ expression.



The phantom assembly shown comprises three adjacent PMMA sections, each 15-cm long and of 32-cm diameter, assembled along the patient table and held together with filler rods in mutually aligned holes. Overall length of the assembled phantom is 45 cm.

Helical translation is more convenient than axial.

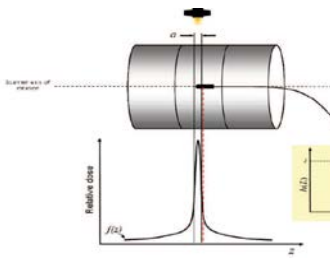
b < l, that is the chamber active length should be higher than the distance between the slices midpoint (possibly b < l/2). Thus the pitch can be computed as p < l/nT that avoids the dose oscillation in the active volume (best: p<l/2nT). For nT= 40 mm and l= 23 mm than p=0.25

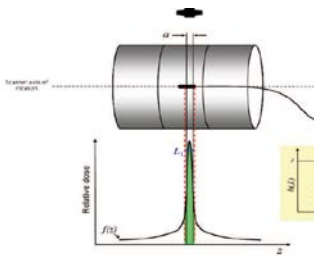
- If a is known then Deq, periphery can be calculated for nT> 40 mm from a value of Deq, periphery for nT <40 using the **equilibrium dose constant**: b < l, that is the chamber active length should be higher than the distance between the slices midpoint.
  - $$\left(\frac{p.nT}{a} D_{eq}{}^{periphery}\right)^{nT<40} = \left(\frac{p.nT}{a} D_{eq}{}^{periphery}\right)^{nT\geq 40}$$

OTHERWISE one can use the relationship:

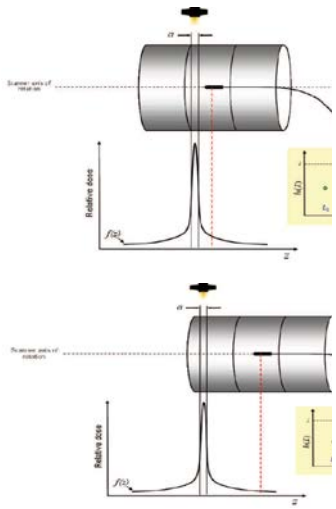
$$\frac{\left(p_1 D_{eq}{}^{periphery}\right)^{nT\geq 40mm}}{\left(p_2 D_{eq}{}^{periphery}\right)^{nT<40mm}} \approx \frac{\left(p_1 D_{eq}{}^{center}\right)^{nT\geq 40mm}}{\left(p_2 D_{eq}{}^{center}\right)^{nT<40mm}} \rightarrow \left(p_1 D_{eq}{}^{periphery}\right)^{nT\geq 40mm} \approx \left(p_2 D_{eq}{}^{periphery}\right)^{nT<40mm} \times \frac{\left(p_1 D_{eq}{}^{center}\right)^{nT\geq 40mm}}{\left(p_2 D_{eq}{}^{center}\right)^{nT<40mm}}$$

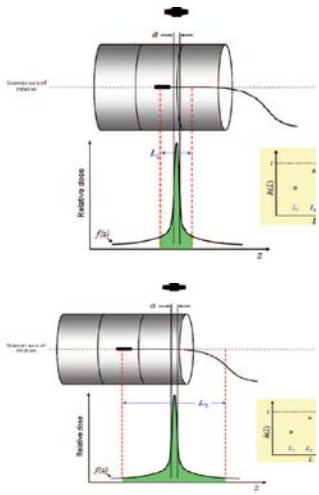
- Practically what is proposed to determine D_L is sketched in the following images:





Ranging from 50 mm to (phantom length –





nT) determine the $D_L(0)$ using the formula proposed by Dixon and Ballard based on the concept of h(L),

MAKE A FITTING of

$$D_L(z) = h(L$$

Then calculate:

$$\hat{D}_{eq} = p.D_{eq}$$

$$\left(\frac{b}{a}\right)D_{eq}$$

- HERE THE FIRST PART IS CONCLUDED WITH THE CALCULATION OF THE PARAMETERS CHARACTERIZING THE REF CONDITION.
- Now retaining *bow-tie filter*$_{ref}$, and *kVp*$_{ref}$ one has to change the other parameters *focal spot size, n, T, a,* and exploiting the $\hat{D}_{eq} = p.D_{eq}$ evaluate the $\hat{D}_{eq}$ for the new combination of parameters with the expression:

$$\hat{D}_{eq}(a) = \left(\frac{a}{n_a T_a}\right) \cdot \left(\frac{b_{ref}}{a_{ref}}\right) \cdot D_{eq,ref}(a_{ref})$$

- A SECOND SET OF DATA IS OBTAINED.
- Now the kVp and Bow tie filter have to be changed and measurements repeated in the new configuration.
- When all the parameters have been calculated for the ionizing chamber inserted in the central cavity of the phantom, REPEAT ALL THE PROCEDURE FOR THE PERIFERAL AXIS DOSE
- CHANGE THE PHANTOM AND REPEAT THE WHOLE PROCEDURE (in the document it is not suggested any particular phantom, anyway a difference should be encountered talking

into account the dimension of a body/adult phantom and that of a child/phantom or a head phantom)

- To calculate the free in air parameter apply the same procedure without the employed PMMA phantom.

At the very end of this procedure different data sets of parameters were employed characterizing the CT equipment in the different scenarios (QC and clinical practice). But it is important to underline that, notwithstanding the difficulty of the procedure and its length, the only (leading) parameter acquired is such Deq related to the particular dose function that "defines" the radiation delivered by that given equipment. The AAPM document is devoted to quality control protocols, even if the whole document is addressed to define a metric available to substitute CTDI and being "more correct" for the physical aspect. The leading parameter influencing the discussion is the queues of the dose profiles that can be larger than the typical CTDI definition 100 mm. For instance a L_{eq} value is introduced (suggested for the same CT equipment) of the order of 400 mm and makes the metrics independent of the scanning length likewise for the concept of DLP/KLP. The only parameter addressed to the delivered dose is the Integral Dose E_{tot} that can be used to calculate the average energy supplied during a CT considering the real weight of the body.

Discussion

At present the AAPM approach seems to be adopted by the "hard core" CT community dedicated to ushering in the change that TG-111 has envisaged. A parallel development is for a CT phantom that could unanimously be agreed on, and AAPM TG-200 (as well as the ICRU) is in progress with such a design. A prototype phantom has been tested in several institutions; it is 60 cm long, 30 cm diameter polyethylene phantom which will allow the principles of TG-111 to be implemented - it's length allows the full assessment of the "rise to equilibrium curve". In addition from the ICRU perspective, this phantom provides a hybrid approach as it will allow the relatively easy measurement of MTF and NPS to assess image quality at the same time dose is measured (Section 3.3.2).

There is a wide agreement that the ideal dose measurement would result in a relatively accurate assessment of organ dose for a standard person and in some cases for a specific person. The methods of BEIR VII can then be used for the assessment of risk if that needs to be computed. Effective dose is not a good measure of an individual's risk, while can be used for a generic person for global purposes. Accordingly, also the "average whole body dose" that can be calculated from the Integral Dose (Etot) introduced by the AAPM is not a good measure for risk assessment, however it correlates with DLP and effective dose.

The new dose metrics are not expected to replace the dose quantities which are used for setting the Diagnostic Reference Levels (DRLs) for patient dose. At least in the short term, it is likely that the traditional quantities CTDIvol and DLP will be used in the measurements of CT scanner output for the setting of DRLs, while these quantities do not really represent patient dose since patient size, age and gender are not included in the calculation. In the USA, the American College of Radiology is launching a dose reporting tool that will be used to compare institutional doses for major CT exams (head, body, chest, etc) against those at other institutions; this is a practical tool that is dynamic and can be adapted to changing reference doses.