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GUEST EDITORIAL

An approach to diagnostic radiology dosimetry

Donald McLean

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In 2003 at the World Congress at Medical Physics in Sydney I heard a talk by František Pernička, late of the International Atomic Energy Agency (IAEA), speaking on an international code of practice for diagnostic radiology (now known as TRS457 [1]). At the time it appeared to me that there was little new or special. Now, over 10 years later, I find I need to review this assumption.

From my perspective much has changed in the last decade. The perception of the importance of diagnostic radiology dosimetry has broadened. Its application now crosses speciality boundaries where technology is posing increasing dosimetric challenges with the needs for standardisation and direction ever present.

Perhaps a good place to start is with the basis of the dosimetry. Unlike dosimetry in radiation therapy which, according to the cover of TRS398 [2], is based on absorbed dose in water, diagnostic radiology dosimetry is based on air kerma [1, 3]. We should accept this basis, as indeed the National Council of Radiation Protection and measurement (NCRP) has in a recent standard on interventional procedures [4] where the units used in that document follow the ICRU report No 74 [3]. Indeed the use of the letter K replacing the letter D is being seen more commonly, for example, kerma area product (KAP) is beginning to replace the former dose area product (DAP) as seen in the new edition of the UK text from Martin and Sutton [5]. The ICRU nomenclature is indeed a little challenging, with the air kerma-area product notated as $P_{K,A}$, similarly air kerma-length product notated as $P_{K,L}$, however this does reinforce an interesting dosimetric principle that is probably not commonly found in radiotherapy.

While radiotherapy dosimetry has a number of parameters designed to monitor dose distributions, a primary dosimetric objective is to determine point doses, which can be related to a severity of radiation damage from a tissue reaction effect. There is often little need for an integral dose metric which could be related to a probability of cancer induction. Typically the reverse is true in diagnostic radiology dosimetry, and hence the usefulness of $P_{K,A}$ generally in diagnostic radiology and $P_{K,L}$ which describes the measurement of a pencil ionisation chamber, as used typically in dental or CT dosimetry. The initial ICRU nomenclature for CT dosimetry [3], was the use of the C_K family to replace the IEC [6] defined CTDI terminology, however the recent ICRU publication on CT dosimetry [7] sees the return of CTDI usage.

At the end of chapter 3 in TRS457 there is found a small equation (3.21) that has had profound effect on my view of

D. McLean (✉)
Canberra Hospital, Canberra 2605, Australia
e-mail: donald.mclean@act.gov.au

dosimetry. It simply relates organ (or tissue) dose to a measured or calculated quantity with correction factor.

$$cf = \frac{\text{Organ or tissue dose}}{\text{Measured or calculated quantity}}$$

with an example that relates organ dose, D_T, to the measured quantity incident air kerma, K_i, thus

$$c_{D_T,K_i} = D_T/K_i$$

The conversion factor here follows the ICRU convention of using suffixes to indicate the two related quantities.[1] One important application here is the determination of skin dose from K_i with conversion factor being the product of the appropriate backscatter factor and the conversion from air kerma to dose in skin (or water). Useful backscatter factors for this are recently found in the literature [8, 9] including those needed for paediatrics [10]. Perhaps a more complex example is found in mammography with the dose to the breast glandular organ, D_G given by

$$D_G = cf\, K_i$$

Here values of *cf* are given in the literature [11–14]. There is good agreement between the values of D_G arrived at using the conversion factors of Wu et al./Boone et al. and Dance et al., notwithstanding the slight differences in the dosimetric models used to generate the conversion factors. One point however that is worth making is the factorial nature of the conversion factors, which include tube voltage, half value layer (HVL), target filter combination, breast thickness and glandularity. Dance et al. have arranged their conversion factor as a product of component conversion factors, thus

$$cf = gcs$$

where g gives the D_G for a breast of glandularity 50 % as a function of HVL and breast thickness, c is the factor that corrects for glandularity, and s is a factor that allows different X-ray spectra and depends on the target filter combination. More recently with the inclusion of digital tomography in mammography, an additional component conversion factor has been added [15], T, which is a function of projection angular range and breast thickness. This neat solution to a complex dosimetric problem points to an approach to deal with emerging (and existing) dosimetric problems in diagnostic radiology.

The IAEA code of practice for diagnostic dosimetry, TRS457, identifies five sets of application specific dosimetric quantities corresponding to the applications of basic radiography, fluoroscopy, CT, mammography and dental radiography. The application of dosimetry to CT, which has been recently reviewed by Kalender [16], is perhaps the area attracting most attention currently in diagnostic radiology. TRS457 recognises the concepts of CTDI, measured with a 100 mm pencil ionization chamber, as defined by the IEC [6]. As indicated by the name, CT dose indicator, the CTDI is defined in air and in specific phantoms and is simply an indicator that can be used for comparative purposes. Further the so called dose length product, DLP, is hence also a dose indicator, which can be converted to a crude measure of effective dose through the use of conversion factors. This process does not take into account the size of the patient, except for some standard paediatric sizes, with the conversion factor assuming different values for different parts of the body scanned. With the advent of multi detector CT (MDCT) equipment, the capability of the pencil ionization chambers to fully capture primary and scatter components within CTDI measurements have been discussed [7, 17–19]. A solution to determine $CTDI_{100}$ measurements for wide beam CT scanning has been advocated initially by the IEC [20].

The problem still remains however about how to relate a dose indicator (in this case CTDI) to a patient dose, such as effective dose, or better to organ doses, to allow risk coefficients to be calculated using recognised coefficients such as those from the BEIR VII report [21]. In the case of CT a good answer appears to be with the use of CT dose software that simulates a wide range of phantom sizes. Examples of such software include Impactdose[2] which can give an estimation of effective dose. For closer patient modelling for patient shapes with outcomes that include organ dose estimates other software is also available [22, 23], ImpactMC[3] which can also be used for cone beam CT (CBCT) applications [24].

The area of CBCT dosimetry is still a work in progress. In addition to the CTDI formalism mentioned above, CBCT can use KAP as the dose indicator as seen in angiographic applications of CBCT [25] and in dental CBCT where direct measurement with TLD is also utilised [26–28].

The effect of size is critical for meaningful personal, or even population, dosimetry for diagnostic radiology [10]. For paediatric dosimetry a reliable indicator of size is required. Often parameters such as age, weight are used by necessity, but may be poor substitutes for dimensional estimates of size [29–32]. It can also be seen that generally the effect of size presents challenges when analysing recorded dose audit data as dose is often increasing exponentially. The relatively low frequency of paediatric examinations and the variety in population size makes

[1] One practical problem with this nomenclature is in the difficulty of writing as many word processing packages have difficulty with two layers of suffixes.

[2] http://www.ct-imaging.de/en/ct-software-e/impactdose-e.html.

[3] http://www.ct-imaging.de/en/ct-software-e/impactmc-e.html.

paediatric dose audit a difficult undertaking [10]. For individual patient dose determination the use of size specific software, such as PCXMC, is invaluable.

Finally we need to consider the newly emerging dose phenomenon that could be called 'DICOM dose'. This is dose indication information mined from DICOM tags that is becoming increasingly accessible. There is a need to ensure that the indicators are properly calibrated, noting that some indicators may have high international tolerances, as is the case for KAP with a 35 % tolerance [33]. It should also be noted that such indicators have very limited value in determining individual dose unless some reliable size indicator is also included. As a population dose indicator it may well have merit although its comparison to diagnostic reference levels (DRLs) may only be relevant if the size profile of the population is a close match to population profile of the study used to develop the DRL.

In conclusion the study of dosimetry in diagnostic radiology has developed rapidly recently, with advances in Monte Carlo software and phantom development that is accessible to the clinical medical physicist. These advances have made it possible to understand and relate the many dose indicators that are used clinically with measures of effective dose and even organ dose in some situations. At the same time a greater understanding of the importance of size in individual dosimetry has informed the use of dosimetry clinically. Further the use of dose indicators to determine good practice through the use of DRLs is developing, along with the use of automated dose assessment, however special care is needed in these applications to avoid erroneous conclusions about clinical practice.

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