Practical patient dosimetry for partial rotation cone beam CT

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Objectives: This work investigates the validity of estimating effective dose for cone beam CT (CBCT) exposures from the weighted CT dose index (CTDI_W) and irradiated length.

Methods: Measurements were made within cylindrical poly(methyl methacrylate) (PMMA) phantoms measuring 14 cm and 28 cm in length and 32 cm in diameter for the 200° DynaCT acquisition on the Siemens Artis zee fluoroscopy unit (Siemens Medical Solutions, Erlangen, Germany). An interpolated average dose was calculated to account for the partial rotation. Organ and effective doses were estimated by modelling projections in the Monte Carlo software programme PCXMC (STUK, Helsinki, Finland).

Results: The CTDI_W was found to closely approximate the interpolated average dose if the positions of the measured doses reflected the X-ray beam rotation. The average dose was found to increase by 8% when the phantom length was increased from 14 to 28 cm. Using the interpolated average dose and the irradiated length for effective dose calculations gave similar values to PCXMC when a double-length (28-cm) CT dose index phantom was irradiated. Simplifying the estimation of effective dose with PCXMC by modelling just 4 projections around the abdomen gave effective doses that were only 7% different to those given when 41 projections were modelled. Calculated doses to key organs within the beam varied by as much as 27%.

Conclusion: Estimating effective dose from the CTDI_W and the irradiated length is sufficiently accurate for CBCT if the chamber positions are considered carefully. A conversion factor can be used only if a single CT dose index phantom is available. The estimation of organ doses requires a large number of modelled projections in PCXMC.

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Using CT in conjunction with a fluoroscopic interventional procedure can provide enhanced anatomical information and greater soft tissue differentiation. The flat panel detectors used widely on fluoroscopy suites and developments in reconstruction algorithms now mean that CT-like images can be obtained by a cone beam CT (CBCT) system fully integrated within a fluoroscopy unit.

As this technology becomes widespread, it is essential to have a measure of the dose to a patient from this type of exposure. For fan beam CT, organ and effective doses may be estimated by measuring the CT dose index (CTDI) in air, and applying a series of scanner-specific conversion factors for the portion of the body irradiated. These factors were calculated by the National Radiological Protection Board (NRPB) [1] using Monte Carlo techniques for an anthropomorphic phantom, based on that of Cristy [2].

A convenient method of applying the NRPB conversion factors is provided by the ImPACT Dosimetry Spreadsheet (ImPACT, London, UK). In addition to the CT scanners originally surveyed by the NRPB, this spreadsheet has matched newer scanners to appropriate factors by matching the ratio of CTDI measured in air and within a poly (methyl methacrylate) (PMMA) phantom. This matching can be done for any scanner and this approach has been used by Sawyer et al [3] for a CBCT

system that rotates 360° around a patient. There are, however, no conversion factors for scanners which perform a partial rotation around the patient.

An alternative method for calculating an approximate effective dose from a CT scan is given in the European Guidelines for Multislice Computed Tomography [4]. This uses weighted CTDI (CTDI_W) and the irradiated scan length. CTDI_W is a weighted average of doses measured at the centre and periphery of a PMMA phantom and is indicative of the average dose within an irradiated slice. For helical scanners, CTDI_W is divided by pitch to give CTDI_{vol} and multiplied by the irradiated length to give the dose-length product (DLP). Effective dose may be estimated from the DLP by applying one of six normalised effective dose per DLP values (E_D) for different body regions.

The calculation for CTDI_W is designed for X-ray tubes that perform a 360° rotation and may not be indicative of the average dose within a slice for a partial tube rotation. In addition, CTDI values are conventionally measured with a pencil dosimeter under the assumption that the collimated X-ray beam and its penumbra are contained within the length of the dosimeter. As this is not the case for CBCT systems, there have been discussions regarding the appropriateness of using CTDI for CBCT dose measurements [5, 6].

Recent work has suggested that dose measurements will be more accurate if a point chamber is used instead of a pencil chamber [7], or if a long pencil chamber (250 mm) is used to capture the entire dose profile [8]. Integrated dose profiles have been compared with measured values of CTDI [3, 5, 6, 8, 9] and all authors

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agree that it is necessary to have an appropriate length of scattering material to contain the full penumbra of the X-ray beam.

For CBCT, an alternative approach for dose calculation is a method commonly used for radiographic and fluoroscopic exposures. The Monte Carlo modelling software PCXMC (STUK, Helsinki, Finland) [10] simulates an X-ray beam by projecting it onto a modified version of the Cristy anthropomorphic mathematical phantom. This gives both organ and effective doses and has recently been used by Wielandts et al [11] for CBCT. Because the beam spectrum and geometry of each exposure is simulated individually, this technique offers a greater degree of accuracy than those developed for conventional CT. However, CBCT is made up of a large number of projections, so this is potentially a time-consuming procedure.

The aim of this study was to determine appropriate methods of estimating organ and effective doses from a partial rotation CBCT acquisition using tools which are readily available. Three methods of determining the average dose within a partially irradiated slice were compared: two using the empirical $CTDI_W$ equation and one using an interpolated average dose calculation. Doses were measured for three different configurations of PMMA phantom and beam width. From this, correction factors were calculated to convert the dose measured in a single PMMA phantom to the dose measured in a longer phantom, and to convert the dose from a thin beam width to the dose from a wide beam width. Effective dose calculations from PCXMC and interpolated average dose measurements were compared, and the number of projections necessary to model the CBCT exposure in PCXMC is considered here in relation to the effect on effective and organ doses.

Method

Equipment

The CBCT acquisition studied was the 8 s DynaCT digital radiography acquisition (8sDR) on a Siemens Artis zee interventional fluoroscopy unit (Siemens Medical Solutions, Erlangen, Germany).

The acquisition consists of 396 fluoroscopy images taken over a 200° rotation around the patient moving from a left anterior oblique (LAO) projection posteriorly to a right anterior oblique (RAO) projection. The acquisition was set up to deliver 0.36 µGy per frame to the imaging plate and no extra copper filtration was selected. The target kV was set to be 70 kV, although pre-acquisition screening determined the actual exposure parameters (kV and mA) required. The uncollimated X-ray beam was measured to be 27 × 36 cm at the imaging plate (craniocaudal × lateral directions, respectively). The field could be collimated in the craniocaudal direction only and the focus-image distance was 120 cm.

Doses were measured with a 100-mm-long 3-cm³ 2025-series Radcal pencil ionisation chamber and electrometer with a calibration traceable to primary standards. The chamber was used with a PMMA CTDI phantom (ImPACT). A single phantom is 14 cm long with a 32-cm diameter. A double-length phantom (28 cm long)

was constructed by fixing two phantoms together. Doses were measured in the centre of the phantom and at the eight outer chamber positions, 15 cm from the centre. The phantoms were placed at the centre of rotation of the tube such that the central chamber position was 60 cm from the imaging plate.

Validating $CTDI_W$

Measurements of $CTDI_W$ were made for three combinations of two phantom lengths (14 and 28 cm) and two X-ray beam collimations (21 and 27 cm at the imaging plate). The phantom/collimation combinations measured were:

- A: a single phantom with the beam collimated to the chamber length;
- B: a double-length phantom with the beam collimated to the chamber length;
- C: a double-length phantom with an uncollimated beam.

For all three combinations, the chamber was placed centrally within the phantom(s) in the craniocaudal direction, and doses were measured at the eight outer chamber positions and the centre chamber position. Collimation was set visually on the display monitor before each exposure. Reproduction of collimation was estimated to be 1 cm in 21 cm for the craniocaudal direction (5%).

Combination A is the preferred testing scenario for routine quality assurance measurements of radiation output; usually only one phantom will be available for use, and it is convenient to transport only one phantom. Combination B is an approximation to an infinite length of PMMA and will include the effect of scatter from the craniocaudal directions. This combination has been shown to most closely represent a true measure of $CTDI_W$ [8]. Combination C more closely mimics patient exposure, including both the craniocaudal scatter caused by the patient body and the larger X-ray field.

The $CTDI_W$ is determined from doses measured from a 360° exposure assuming that there is a linear decrease in dose between the periphery and the centre of the phantom [12]. For a beam width W less than the length of the pencil chamber L , it is given by the empirical equation:

$$CTDI_W = (1/3D_{\text{centre}} + 2/3D_{\text{periphery}})L/W \quad (1)$$

where D_{centre} is the dose measured in the centre of the phantom and $D_{\text{periphery}}$ is the average of the doses measured at the outer chamber positions of the phantom. $D_{\text{periphery}}$ is usually calculated using four dose measurements, typically taken at the north, east, south and west positions in the phantom.

When the beam width is greater than the length of the chamber, W is given the value of L [9] and Equation (1) simplifies to:

$$CTDI_W = (1/3D_{\text{centre}} + 2/3D_{\text{periphery}}) \quad (2)$$

To investigate whether this relation still holds for a partial irradiation over 200°, this empirical $CTDI_W$ value was compared with an interpolated average dose within

Table 1. Exposure details used in PCXMC (STUK, Helsinki, Finland) to model a DynaCT (Siemens Medical Solutions, Erlangen, Germany) examination of the abdomen

Measurement scenario	Image dimensions (cm)	kV	DAP (mGy cm ²)	
			Per projection	Total
B. Double phantom, collimated beam	21 × 36	124	1301.2	53347
C. Double phantom, uncollimated beam	27 × 36	121	1721.9	70596

DAP, dose-area product.

a slice. Assuming the same linear decrease between the periphery and the centre of the phantom, the phantom was divided into a polar grid of 1-cm radius intervals and $\pi/8$ angular intervals. Doses were then interpolated between the eight periphery dose values and the centre dose value to give doses at each grid position. The interpolated average dose in the slice (D_{ave}) was found for each phantom/collimation combination and compared with two empirical CTDI_W values. The first (CTDI_W^{ON}) was calculated using the four 'on-axis' periphery doses measured at the north, east, south and west chamber positions in the phantom. CTDI_W^{OFF} was calculated using the four 'off-axis' periphery doses measured at the northeast, southeast, southwest and northwest chamber positions.

Effective dose calculations: PCXMC

The effective dose from a DynaCT 8sDR examination of the abdomen was modelled with the PCXMC 2.0 dose calculation software. The exposure was initially modelled as 41 separate projections at 5° intervals around the PCXMC phantom covering a 200° rotation. An anode angle of 12° and tube filtration of 2.5 mm Al were used as tube parameters and the focus-image distance was 120 cm for all projections.

Because the mathematical phantom in PCXMC was elliptical, it was necessary to calculate the distance from the entry surface to the image plate for each projection to determine the dimensions of the beam as it entered the phantom. The phantom was chosen to be the standard adult patient in PCXMC, weighing 73.2 kg and measuring 178.6 cm in height.

The PCXMC software requires a measure of exposure to calculate the effective dose from each projection. The dose-area product (DAP) of the entire acquisition was used, divided by the number of projections used to simulate the exposure. The final kV reported by the unit was used for all projections. Wielandts et al [11] used frame-by-frame kV and mAs values extracted from the X-ray unit. They demonstrated that tube power remains constant throughout the run, so assuming constant kV and DAP per frame was a reasonable approximation. Exposure conditions for phantom/collimation combinations B and C were modelled in PCXMC. Combination A was expected to give the same result as B because the kV and DAP values were similar and the different phantom lengths had no effect on the PCXMC calculations. The image dimensions, DAP values and kV values used are shown in Table 1.

Because the effective dose varies with the organs irradiated, the examination was simulated in PCXMC at three locations within the abdomen (Figure 1). The upper model covered the liver, stomach and spine;

the middle covered the intestines and upper pelvic bones; and the lower covered the lower intestines and the reproductive organs. PCXMC 2.0 calculates the effective dose using the updated tissue-weighting factors issued in International Commission on Radiological Protection (ICRP) publication 103 [13] as well as the weighting factors given in ICRP publication 60 [14].

In addition to modelling the DynaCT acquisition with 41 projections, the uncollimated beam acquisition (combination C) was modelled with a decreasing number of projections. The angle between projections was increased in 5° steps, eventually modelling the acquisition as four projections 55° apart. All three locations within the abdomen were simulated with the reduced projections.

Because effective dose is calculated from organ doses, the effect of reducing the number of modelled projections on individual organ doses was examined in addition to the final effective dose value.

Effective dose calculations: European Guidelines for Multislice Computed Tomography

For the abdomen and pelvis region, the effective dose conversion coefficient provided by the 2004 European Guidelines for Multislice Computed Tomography is $E_D=0.017 \text{ mSv mGy}^{-1} \text{ cm}^{-1}$.

Figure 1. Left, screenshot of the mathematical phantom in PCXMC (STUK, Helsinki, Finland) with the location of the three abdomen exposures shown. Right, organs irradiated in each of the three abdomen exposures. From the upper abdomen the primary organs shown are: liver, stomach and spine; intestines and upper pelvic bones; lower intestines and reproductive organs.

The interpolated average doses from the three phantom/collimation combinations were used with this coefficient to provide an estimate of effective dose from the DynaCT acquisition:

$$\text{Effective dose} = E_D \times D_{\text{ave}} \times \text{irradiated length} \quad (3)$$

The irradiated length was taken to be the length of the X-ray beam at the centre of the phantom. The effective dose conversion coefficient has been calculated on the basis of the ICRP 60 tissue-weighting factors so the doses were compared with the equivalent effective dose values calculated using PCXMC.

Results

Validating CTDI_W

D_{ave} , CTDI_W^{ON} and CTDI_W^{OFF} are given in Table 2 for the three phantom/collimation combinations. The third column gives the correction factors between the interpolated average doses measured in the different phantom/collimation combinations.

When the beam remained collimated and the phantom length was doubled, the D_{ave} increased by 8%. When two phantoms were used and the X-ray beam changed from collimated to uncollimated (increasing the beam length at the image plate by 6 cm), there was a further increase of 8%.

The fourth and fifth columns in Table 2 show the percentage differences between the interpolated average dose in a slice and the average doses calculated from the empirical CTDI_W equations. CTDI_W^{ON} was an average of 13% greater than the interpolated measure of average dose and CTDI_W^{OFF} was an average of 5% greater.

Effective dose

Table 3 gives the effective doses from the phantom exposures calculated using the effective dose conversion coefficients from European guidelines (column 2), and using PCXMC (columns 3–5). There are three sets of results from PCXMC because exposures of three sections of the abdomen were simulated. The doses calculated from the European conversion coefficients use the interpolated average dose in a slice rather than the empirical CTDI_W values. Using CTDI_W^{ON} or CTDI_W^{OFF} will increase the estimated effective dose by 13% and 5%, respectively.

Figure 2 shows how the effective dose from PCXMC varies with the number of modelled projections. The model used the exposure factors from the uncollimated beam acquisition (combination C) and Figure 2 shows the results from the middle abdomen exposure, covering the intestines and upper pelvic bones. When compared with the values found using 41 projections, effective doses were up to 7% different for the upper abdomen region, and up to 2% different for the lower and middle abdomen regions.

Organ doses

Figure 3 shows the change in calculated organ doses for a selection of organs as the number of projections used to model the DynaCT exposure is varied. The results shown are for the uncollimated beam exposure covering the middle abdomen area (intestines and upper pelvic bones). Trends are similar for the upper and lower abdomen exposures. The organ doses are expressed as the percentage difference from the dose obtained using 41 projections. Four organs within the primary beam are shown: liver, stomach, small intestines and pelvis. Two organs far from the primary beam are also shown: lower leg bones and salivary glands. Table 4 summarises key organ doses calculated with 41 projections for all 3 locations in the abdomen, and gives the maximum percentage difference seen from these numbers when the number of projections is reduced.

Discussion

For the phantom/collimation combinations studied here the empirical equation for CTDI_W overestimated the interpolated average dose by an average of 5% if the 'off-axis' doses were used, or 13% if the 'on-axis' doses were used. The difference in the two CTDI_W values was due to the chamber positions and the partial rotation of the tube. This exposure was over a 200° rotation moving from an LAO projection posteriorly to an RAO projection. Using the 'off-axis' chamber positions better reflects the area irradiated because the primary X-ray beam never enters the 'north' side of the phantom. To calculate the average dose within a slice for this type of exposure with only four measurements, it is therefore most accurate to use the doses measured at the northeast, southeast, southwest and northwest chamber positions.

An equivalent comparison of interpolated average dose and empirical CTDI_W for a fan beam Mx8000 CT scanner (Philips Healthcare, Best, The Netherlands) was

Table 2. Interpolated average doses (D_{ave}) and weighted CT dose index (CTDI_W) values for the three phantom/collimation combinations

Measurement scenario	D_{ave} (mGy)		% difference between CTDI _W and D_{ave}	
	Value	Normalised to A	CTDI _W ^{ON}	CTDI _W ^{OFF}
A. Single phantom, collimated beam	53.9	1.00	+12	+6
B. Double phantom, collimated beam	58.2	1.08	+14	+4
C. Double phantom, uncollimated beam	62.5	1.16	+12	+5

The third column gives the interpolated average dose measurements normalised to that of a single phantom. CTDI_W^{ON} was calculated using doses measured at the 'on-axis' chamber positions in the CTDI phantom. CTDI_W^{OFF} was calculated using doses measured in the 'off-axis' chamber positions.

Table 3. Effective doses from the three phantom/collimation combinations

Measurement scenario	Effective dose (mSv)			
	Using European criteria and D_{ave}	Using PCXMC with ICRP 60 (ICRP 103)		
		Lower abdomen	Middle abdomen	Upper abdomen
A. Single phantom, collimated beam	9	12 (9)	10 (9)	13 (12)
B. Double phantom, collimated beam	11			
C. Double phantom, uncollimated beam	15	15 (12)	14 (13)	16 (16)

European criteria doses are calculated using ICRP 60 tissue-weighting factors. PCXMC (STUK, Finland, Helsinki) effective doses are given for both ICRP 60 and ICRP 103 tissue-weighting factors. Combination A will give the same result as combination B in PCXMC so only one combination was modelled (B).

carried out. The interpolated average dose in a slice was calculated using four doses measured 'on-axis'; this is commonly expected to be sufficient for a 360° exposure. The calculated $CTDI_W$ in this case overestimated the interpolated average dose by 6%. This is comparable to the 'off-axis' overestimation for the 200° rotation. The overestimation of the empirical $CTDI_W$ is not corrected for when estimating doses from a fan beam CT scanner so it did not seem necessary to correct the CBCT result either.

When interpolating the measured doses to get an average dose in a slice, doses were assumed to decrease linearly from the outside to the centre of the phantom. This is the case for CT beams because opposing X-ray beams being attenuated exponentially will result in a linear decrease towards the centre of a phantom. This will not be valid for the entire phantom when considering a 200° rotation, however, as there will be a fraction without opposing beams. In addition, the fluoroscopic beam will have a broader energy spectrum than a CT

system, so the assumption of strict exponential absorption will be less appropriate.

Ideally, patient dose measurements would be made with a double-length phantom as this has been shown in the literature to be long enough to accurately capture the dose profile of a 200-mm collimated beam [8]. This is not always practical, owing to the size and availability of the phantoms. A routine measure of patient dose for this unit can instead be made in a single phantom and the result multiplied by a correction factor of 1.08. This will give the $CTDI_W$ value for a collimated patient exposure without the need to use two phantoms. A multiplication of 1.16 will approximate to an uncollimated beam exposure if this better represents the clinical exposure.

The effect of irradiating different organs can be seen by comparing the effective doses from the three PCXMC simulations covering different abdominal regions.

The difference in doses can be explained by the different organ distributions and weighting factors. The upper and lower abdominal regions contain many organs with individual weighting factors, and the middle region mostly covers the small intestines, which are classed as one of 13 "remainder" organs in ICRP 103.

When modelling any exposure in PCXMC, the accuracy of the result will depend on reproducing the position and size of the X-ray beam on the patient. Limitations of PCXMC may result in a mismatch of irradiated organs, e.g. if the field size is correct, the irradiated organ volumes may be incorrect.

For ICRP 60 tissue-weighting factors, effective doses calculated using the European conversion coefficient were all similar to those calculated in PCXMC, and were best matched for the uncollimated exposure of the double phantom. Using the European coefficient for the single phantom exposure gave up to a 30% lower effective dose than the PCXMC method. To use the European conversion coefficient method with a single phantom exposure it appears necessary to use the correction factors given above to change the $CTDI_W$ value into one for a double phantom exposure before a good comparison with PCXMC is seen. This difference will be due to the effect of scatter in the cranial and caudal directions being included in the PCXMC dose calculation method. It should also be noted that there is only one European conversion coefficient for the entire abdomen region. Effective dose conversion coefficients have been calculated for smaller irradiated areas within the body [15]. This approach is more sophisticated owing to its better resolution; however, this degree of

Figure 2. The variation in effective dose with the number of projections modelled in PCXMC (STUK, Finland, Helsinki). Results are for a middle abdomen exposure, covering the intestines and the upper pelvis bones. The percentage difference from the effective dose calculated with 41 projections is plotted (14 mSv). Effective doses are calculated using ICRP 60 tissue-weighting factors.

Figure 3. The variation in organ dose with the number of projections modelled in PCXMC (STUK, Finland, Helsinki). Results are for a middle abdomen exposure, covering the intestines and the upper pelvis bones. The doses from 41 projections are given in the graph headers and the percentage differences from these are plotted. (a–d) Liver, stomach, small intestines and pelvis are within the primary beam. (e,f) Lower leg bones and salivary glands receive only scattered radiation.

accuracy is not necessarily required with a cone-shaped X-ray beam, in which tight collimation is not used clinically.

Comparison of effective doses calculated using the European conversion coefficient with those from PCXMC using ICRP 103 tissue-weighting factors gave a maximum difference of 33%. However, this cannot be a maximum for other areas of the body, owing to the changes in tissue-weighting factors between 60 and ICRP 103.

ICRP 103 recommends that estimates of risk to the patient should be based on doses to organs rather than effective dose [13]. When the number of modelled projections in PCXMC was reduced, the organ doses were found to vary significantly from those calculated with 41 projections. The organs within the primary beam showed variations of up to 27% (e.g. liver, stomach, small intestines and pelvis). The doses to organs receiving only scattered radiation (so receiving fewer photons) can vary

Table 4. Doses to key organs in the abdomen using exposure factors from an uncollimated exposure on a double phantom (phantom/collimation combination C)

Organ	Organ dose (mGy) (maximum variation in dose [%])		
	Lower abdomen	Middle abdomen	Upper abdomen
Liver	2 (8)	11 (13)	49 (13)
Stomach	3 (12)	13 (21)	38 (27)
Small intestines	38 (1)	45 (1)	18 (3)
Pelvis	196 (10)	150 (11)	10 (6)

The dose from modelling with 41 projections on 3 different abdomen locations is given. The maximum percentage change from these values when projection numbers are reduced down to four projections is given in brackets.

by much more (e.g. lower legs 260% and salivary glands 95%). Although these differences appear large, the absolute dose values are much lower than for organs within the primary beam.

For all of the organs presented here, the organ dose appears to converge towards the 41 projection value when more than 10 projections are modelled. Using 10 projections will give an accuracy of >5% for organs in the primary beam but, ultimately, the desired accuracy should decide the number of projections modelled.

Conclusions

We have investigated the 8 s cone beam DynaCT acquisition on a Siemens Artis zee fluoroscopy unit. We have found that a routine measurement of patient dose can be made on this unit by using one CTDI phantom and measuring four doses in the 'off-axis' chamber positions. The CTDI_W value from these doses is only 5% greater than the interpolated average dose within a slice and can be corrected for if necessary. CTDI_W can be used to calculate effective dose using published conversion coefficients (for ICRP 60 tissue-weighting factors). However, to give similar results to those achieved with Monte Carlo modelling, it is important to correct for a limited phantom length. For this unit, this requires increasing the average dose by 8% if modelling a collimated beam or by 16% for an uncollimated beam. To accurately calculate organ doses from an abdomen exposure, a large number of projections need to be modelled in PCXMC. However, our results for the abdomen indicate that an adequate estimate of effective dose may be calculated from just four modelled projections, saving considerable time and introducing an error of up to 7%.

References

1. Jones DG, Shrimpton PC. Survey of CT practice in the UK. Part 3. Normalised organ doses calculated using Monte Carlo techniques. London, UK Her Majesty's Stationery Office, 1991.
2. Cristy M, Eckerman KF. Specific absorbed fractions of energy at various ages from internal photon sources. ORNL/TM-8381. Oak Ridge, TN: Oak Ridge National Laboratory, 1987.
3. Sawyer LJ, Whittle SA, Matthews ES, Starritt HC, Jupp TP. Estimation of organ and effective doses resulting from cone beam CT imaging for radiotherapy treatment planning. *Br J Radiol* 2009;82:577–84.
4. Bongartz G, Golding SJ, Jurik AG, Leonardi M, van Persijn van Meerten E, Rodriguez R, et al. European guidelines for multislice computed tomography. Luxembourg: European Commission, 2004.
5. Boone JM. The trouble with CTDI₁₀₀. *Med Phys* 2007;34:1364–71.
6. Mori S, Masahiro E, Nishizawa K, Tsunoo T, Aoyama T, Fujiwara H, et al. Enlarged longitudinal dose profiles in cone-beam CT and the need for modified dosimetry. *Med Phys* 2005;32:1061–9.
7. Fahrig R, Dixon R, Payne T, Morin RL, Ganguly A, Strobel N. Dose and image quality for a cone-beam CT system. *Med Phys* 2006;33:4541–50.
8. Kyriakou Y, Deak P, Langner O, Kalender WA. Concepts for dose determination in flat-detector CT. *Phys Med Biol* 2008;53:3551–66.
9. Amer A, Marchant T, Sykes J, Czajka J, Moore C. Imaging doses from the Elekta Synergy X-ray cone beam CT system. *Br J Radiol* 2007;80:476–82.
10. Tapiovaara M, Lakkisto M, Servomaa A. PCXMC: a PC-based Monte Carlo program for calculating patient doses in medical X-ray examinations. Report STUK-A139. 2nd edn. Helsinki, Finland: Finnish Centre for Radiation and Nuclear Safety, 2002.
11. Wielandts J-Y, Smans K, Ector J, De Buck S, Heidbuchel H, Bosmans H. Effective dose analysis of three-dimensional rotational angiography during catheter ablation procedures. *Phys Med Biol* 2010;55:563–79.
12. Leitz W, Axelsson B, Szendro G. Computed tomography dose assessment: a practical approach. *Radiat Prot Dosim* 1995;57:377–80.
13. International Commission on Radiological Protection. Recommendations of the International Commission on Radiological Protection. ICRP Publication 103. Annals of the ICRP, vol. 37/2-4. Oxford, UK: Elsevier, 2007.
14. International Commission on Radiological Protection. 1990 Recommendations of the International Commission on Radiological Protection. ICRP Publication 60. Annals of the ICRP, vol. 21. Oxford, UK: Pergamon Press, 1991.
15. Huda W, Ogden KM, Khorasani MR. Converting dose-length product to effective dose at CT. *Radiology* 2008;248:995–1003.