

Dosimetry of CBCT: methods, doses and clinical consequences

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Abstract. The use of Cone beam CT (CBCT) systems for Image Guided Radiotherapy is rapidly expanding in the developed world. With its use comes concern for the increased risks of additional radiation exposure. Quantification of the imaging dose is necessary in order to report, optimise and justify CBCT exposures. This article reviews the current methods of dose measurement and calculation including dose measurements in cylindrical phantoms, use of point dosimeters in anthropomorphic phantoms, calculation of dose using mathematical phantoms and calculation of individualised patient dose using Monte Carlo and model based techniques. Typical doses from commercial systems are reported and the clinical consequences, both risks and benefits, of using CBCT based IGRT reviewed briefly.

1. Introduction

Cone beam CT (CBCT) systems integrated into a radiotherapy treatment room (Linac) have become readily available within radiotherapy departments since Jaffray *et al* first published the concept in 1998 and becoming commercially available in 2005. The CBCT system enables a sequence of 2D radiographic projection images to be acquired from a kV source and flat panel detector imaging system as it rotates around the patient either before treatment or more recently during treatment [1]. These projection images can be reconstructed to produce a 3D "volumetric" image which is similar, but not of the same quality, as a conventional fan beam CT scanner. The reconstructed CBCT images can be used to correct patient position prior to treatment [2,3] or as a basis to adapt [4–6] the treatment plan to the changing anatomy of the patient during the course of their radiotherapy. This ability to visualize the target anatomy and nearby organs at risk at the point of treatment delivery and to subsequently make corrections either to the patient position or the dose delivered is often termed image guided radiotherapy (IGRT) [7]. The two main commercial radiotherapy CBCT systems from Varian and Elekta are the On-Board Imager (OBI) and Synergy system (XVI, X-ray volumetric imager) respectively.

The introduction of CBCT for IGRT proceeded initially, at least in some countries, with little regard for the extra radiation dose delivered by the imaging system, known as concomitant dose. The generalized justification for this was that the benefits of IGRT will outweigh the increased risks from the CBCT dose. However, more recently there has been increased interest in the concomitant imaging dose. This may be partly driven by a recent campaign aimed in particular at pediatric practice, to apply principles of ALARP – as low as reasonably practicable (Alliance for Radiation Safety in Pediatric Imaging, <http://www.pedrad.org>). There have also been a couple of publications that have made alarming statements about the magnitude of CBCT doses. For example Spezi *et al* found for 40 image fractions of a head and neck using the manufacturers image acquisition protocol that



“the mean concomitant dose in the beam-bordering region accumulated during the RT course from daily use of CBCT (assuming a total of 40 scans) could be in the range 1-2 Gy, with maximal doses in the range of 3-7 Gy.”[8]

With such doses there is rightly cause for concern e.g where the imaging dose increases the dose to the spinal cord for plans where intensity modulated radiotherapy (IMRT) has optimized the dose distribution to near cord tolerance. In such cases it is not easy to know whether the concomitant imaging dose is justified. However, in Stereotactic Ablative Radiotherapy (SABR) treatments of the lung and other anatomical sites such as prostate and spine, where the number of fractions is low (typically 3-8 fractions) and where the effect of geometric miss is much greater, the use of CBCT imaging is clearly justified. Also, for SABR, the total concomitant dose is low because of the reduced number of CBCT scans required. Justification of imaging dose, mandatory in UK legislation [9] and many other countries, is clearly required if CBCT is to be used safely in IGRT. In order to justify the IGRT dose, knowledge of the risks associated with the concomitant imaging, e.g risk of induction of secondary malignancy need to be balanced against the benefits, such as improved treatment outcome, whether that be increase in overall survival or local control or reduced toxicity and side effects both acute and long term. At the present time there is great uncertainty in the risks of secondary cancer induction from radiotherapy [10,11]. There is also very little published evidence on the benefits of IGRT protocols although these are likely to gradually emerge as results of clinical trials are published.

Of course justification of concomitant imaging dose is easier if it is lower and there are many publications which address the optimization of image acquisition, i.e reduction of imaging dose whilst maintaining sufficient image quality to perform the task of IGRT, which is, inherently, one of image registration [12]. The methods employed range from simply reducing the exposure, employing bow tie filters or “zonal” filters [13,14] reducing the number of projection angles used in reconstruction [12,15] and improving the reconstruction algorithms [16,17].

In order to minimize and justify the concomitant imaging dose we need to be able to measure it. The aim of this article is to review the methods employed and the progress to date in the measurement of concomitant imaging dose from CBCT.

2. Review of CBCT dose measurement methods

2.1. Measurement of CBCT dose in cylindrical phantoms

To understand dose measurement in CBCT it is worth taking a brief historical view of dose measurement in conventional fan beam CT (FBCT). The established method has been to measure the Computed Tomography Dose Index (CTDI) [18,19]. This is measured using a 100mm long cylindrical ionisation chamber in a cylindrical phantom. The $CTDI_{100}$, defined in equation (1), measures the dose to air over a volume which encapsulates the slice width plus the tails of the fan beam profile either side. This is performed for a single axial (i.e. not helical) rotation of the x-ray tube. For a narrow slice (≤ 1 cm) a 100mm long chamber is sufficient to capture enough of the profile without significant loss of accuracy. Capturing the entire profile, including the tails due to scatter, is equivalent to measuring the dose for a spiral acquisition,

$$CTDI_{100} = \frac{1}{nT} \int_{-50\text{mm}}^{50\text{mm}} D(z) dz \quad (1)$$

where, n is the number of detector rows and T is the thickness of each row (mm).

Although there are some variants of the CTDI measurement, the $CTDI_{100}$ is generally accepted as a standard [20] and is the standard adopted by the IEC [21].

To estimate the average dose in the axial plane $CTDI_w$ equation (2) is calculated from the average of measurements at the centre and the periphery of the phantom.

$$CTDI_w = \frac{1}{3} CTDI_{center} + \frac{2}{3} CTDI_{periphery} \quad (2)$$

$CTDI_w$ alone only tells us what the average dose in a plane would be given a series of contiguous axial scans (stepped table feed) or helical scans with a pitch of unity. For a helical scan with pitch not equal to unity, volume CTDI ($CTDI_{vol}$) can be calculated using equation (3). The $CTDI_{vol}$ can be multiplied by the scan length to give the Dose Length Product (DLP) as in equation (4) in order to estimate the total absorbed dose, which can then be related to the stochastic effects of radiation exposure and used as a first order estimate in public health monitoring.

$$CTDI_{vol} = CTDI_w / Pitch \text{ (mGy)} \quad (3)$$

Where pitch is the table increment per revolution as a fraction of the detector width (nT)

$$DLP = CTDI_{vol} \times L, \text{ where } L \text{ is the scan length (mGy.mm)} \quad (4)$$

The CTDI concept starts to break down with increasing cone angle for a number of reasons; 1) the dose profile along the central axis becomes non-uniform, 2) the tails of the dose profile extend well beyond the 10cm chamber length, 3) the phantom is not sufficient to capture the entire beam width and scatter and 4) the weightings of central and peripheral dose in $CTDI_w$ are not always a good estimate of the average dose across the volume.

In the publications arising soon after the introduction of CBCT in radiotherapy practise these problems were largely ignored [15, 22, 23] and researchers employed “CTDI like” techniques to measure the CBCT dose. They continued to use a 10cm chamber in a cylindrical PMMA phantom with the same diameters as the body and head CTDI phantoms (32cm and 16cm) respectively but lengthened often by placing two or more CTDI phantoms end to end. In recognition that this only measured an average of a central 10cm portion of the central axis, Amer *et al* gave this measurement the term Cone Beam Dose Index (CDBI). To account for the beam length of 25 cm being longer than the standard 15cm CTDI phantom Amer *et al* added a 15cm thicknesses of PMMA at each end of the phantom.

One method to ensure that the entire dose profile is acquired, at least for Cone angles typical of diagnostic CT scanners, is to measure the dose at the centre of a lengthened CTDI type phantom using a 300mm long cylindrical ion chamber [24].

An alternative is to measure the point dose using a 0.6 cm³ Farmer type chamber, as suggested by Fahrig *et al* [25], in cylindrical PMMA phantoms [26-28]. Song *et al* measured the dose at the centre of two CTDI type (body and head) phantoms placed end to end and gave this measurement the term CBCTDI [27]. Sykes *et al* used the same technique to measure the dose for both the Elekta Synergy and Varian OBI systems [28]. Typical doses for the two systems, taken from Sykes *et al* are given in Table 1.

The equivalence of measuring the dose at the centre of a helical scan of length, L $D_L(z = 0)$ and $CTDI_L$ the integral dose for a single static slice between $-L/2$ and $L/2$ was demonstrated by Dixon *et al* [29]. Furthermore, they demonstrated the equivalence of $D_L(z = 0)$ measured at the centre of a helical scan (scan length = L and pitch = 1) and a CBCT scan (aperture a) where $a = L$. This was confirmed experimentally by Mori *et al* who used a photodiode stepped through the beam to measure profiles of various beam widths [30]. They showed that dose profiles acquired along the central axis of both helical CT and CBCT scans are equivalent. An alternative to the stepping diode which would require many repeat CBCT scans is the use of a CT dose profiler (RTI Electronics AB, Mölndal, Sweden) as demonstrated by Palm *et al* [31].

Current recommendations for dose measurements published by the International Atomic Energy Authority (IAEA) and American Association of Medical Physics (AAPM) are divergent [32, 33].

The IAEA, have followed the IEC 60601-2-44 report [34] which is a pragmatic approach that can be performed with current dosimetry equipment. They note that even for a 10mm wide beam the $CTDI_{100}$, measured at the centre of the phantom, only collects 82% and 63% of the dose for the head and body phantoms respectively [35]. This illustrates that the $CTDI_{100}$ was never as accurate as one might desire with doses under-estimated for long scan lengths and over-estimated for short scan lengths. However, the $CTDI_{100}$ accuracy stayed constant for beam widths between 10mm and 40mm and only decreased significantly for beam widths greater than 40mm. The IAEA recommend a two tier approach to the $CTDI_{100}$ with measurement for beam widths of less than 40mm following the existing method but for those greater than 40mm they exploit equation (5) which states that the CTDI for a beam width greater than 40mm is related to the CTDI for a beam width less than 40mm by the ratio of the $CTDI_{free-in-air}$ at the two beam widths.

$$CTDI_{100,(N \times T) > 40} = CTDI_{100,ref} \times \left(\frac{CTDI_{free-in-air,N \times T}}{CTDI_{free-in-air,ref}} \right)$$

where, $CTDI_{100,ref}$ is the $CTDI_{100}$ measured in a phantom for the reference beam of $(N \times T)_{ref}$ using an integration of 100mm, N = number of detector rows and T is the thickness of a single detector row and where $(N \times T)_{ref}$ is typically 20mm, $CTDI_{free-in-air,N \times T}$ is the $CTDI_{free-in-air}$ for a beam width of $N \times T$, and $CTDI_{free-in-air,ref}$ is the $CTDI_{free-in-air}$ for the reference beam width. (5)

Table 1: Cone Beam dose measurements (similar to $CTDI_w$) for standard imaging protocols on the Varian OBI and Elekta Synergy CBCT systems published in the UK Centre of Evidence Based Purchasing report [28]. Manufacturers recommended protocol settings may change over time based on the introduction of new technology or feedback from customers.

Varian OBI Imaging Protocol	Dose (mGy)	Elekta Synergy Imaging Protocol	Dose (mGy)
Low Dose Head	2.8	Low Dose Head	1.4
Standard Dose Head	5.6	Medium Dose Head	5.4
High Quality Head	27.8	High Dose Head	9.4
Pelvis	24.9	Pelvis M10	15.3
Pelvis Spotlight	20.2	Pelvis M15	12.5
		Pelvis M20	13.7

The measurement of $CTDI_{free-in-air,ref}$ is itself measured in two tiers with a single chamber (100mm length) position being used for beam widths less than 60mm and using two or three positions, each stepped by 100mm to cover beam widths larger than 60mm.

In the AAPM report [33], task group 111 present the theoretical underpinnings of measuring dose in axial or helical fan- or cone-beam CT with table translation and that in stationary-phantom cone beam CT. For axial or helical scanning they note that there is an equilibrium dose constant which is independent of the collimation or the pitch. The dose for any particular scan can then be determined as the product of the equilibrium dose constant and a factor (pnT/a) where p is the pitch, nT the total width of the detector i.e n rows of width T and a is the width of collimation. This considerably reduces

the number of measurements that need to be made as long as the collimation, a , is known. The equilibrium dose constant is the dose measured at the central scan plane ($z = 0$) for a scan of length L_{eq} and with pitch $p=a/nT$. L_{eq} has to be sufficiently large that further increment of L does not significantly increase the measured dose.

Since the 10cm long chamber does not cover the entire beam profile, for wide angle CBCT the 10cm chamber can underestimate the dose by 2-5% [36]. This is one reason why the AAPM TG-111 methods are based on measuring the point dose (e.g using a Farmer type chamber) in a geometrical phantom (typically cylindrical) that is sufficiently long to provide full scatter conditions for the irradiated scan length.

AAPM TG-111 also introduces the concepts of integral dose (E_{tot}) and planar average equilibrium dose ($\overline{D_{eq}}$) and show that $E_{tot} = \rho\pi R^2 L \overline{D_{eq}}$ where R is the radius of the volume, L is the scan length and ρ is the mass density of the phantom. The integral dose,

“serves as a simplified indicator of patient risk: the presumption is that cancer risk increases the larger the dose and irradiation volume containing radiosensitive tissue.”[33]

The practicalities of measuring $\overline{D_{eq}}$, the average dose over the scan plane are discussed briefly. Recognising that when $D_{eq}(r) = A + Br^2$ i.e $D_{eq}(r)$ has a parabolic form, then measurement of $D_{eq}(r)$ at two points such as the centre and periphery of the phantom as conventionally measured for $CTDI_{vol}$, would be reasonable. Note, however that this leads to $E_{tot} = \frac{1}{2} \cdot D(r=0) + \frac{1}{2} \cdot D(r=R-1)$ instead of the more commonly used formula $CTDI_{vol} = \frac{1}{3} \cdot CTDI(r=0) + \frac{2}{3} \cdot CTDI(r=R-1)$. Recognising that $D_{eq}(r)$ does not always follow a parabolic form AAPM TG-111 note that more detailed measurement or the use of Monte Carlo modelling might be required. This would be the case for the large fields of view of the Elekta Synergy and Varian OBI and also the medium field of view of the Elekta Synergy system where the detector panel is shifted laterally to extend the field of view. This creates a central cylinder which is exposed from all 360° whilst material in the remaining volume is only exposed from 180°.

AAPM TG-111 note that measuring the free-in-air dose equilibrium pitch product, is an important measurement to make at commissioning as it can be used for quality assurance purposes to assess constancy of exposure and can be used to infer the equilibrium dose measured in a phantom given a scanner with the same phantom factor (ratio of dose equilibrium in phantom to free-in-air dose equilibrium).

While AAPM TG-111 presents the theoretical underpinnings of the measurement of CBCT dose, it does not offer much in the way of standardisation of CBCT dose measurement. The report suggests but does not dictate the use of a Farmer type chamber and it discusses various phantom designs with various dimensions and cross-sectional shapes (circle or ellipse) and different materials but does not make any recommendations for a standard phantom and chamber as with the CTDI concept. The methods used by Islam *et al*, Song *et al* and Sykes *et al* [26-28] are in many ways closely aligned with TG-111.

2.2. Calculation and measurement of population based organ dose and effective dose from CBCT imaging

The $CTDI_{vol}$ measurement is an estimate of the average dose in the central axial plane of the scan and is typically calculated as one third of the central dose and two thirds of the peripheral dose. This is independent of the scan length and therefore does not relate to the total dose to the patient and any risk of radiation induced malignancy. A commonly used and very simple method to relate $CTDI_{vol}$ to total or examination dose is to multiply by the length of the scan. This is known as the dose length product (DLP).

For a more accurate assessment of radiation risk, the dose to individual organs and their respective organ sensitivities are needed. The effective dose (7), measured in units of Sieverts (Sv), is a summation of tissue equivalent doses (6) and tissue-specific weighting factors defined by ICRP 103 [37]. The effective dose can be related to radiation risk using, for example, data published on the Biological Effects of Ionizing Radiation (BEIR) by the National Academies Concerning Radiations Health Risks [38].

Equivalent dose for tissue/organ T (Sieverts)

$$H_T = \sum_R W_R \cdot D_{T,R} \quad (6)$$

where W_R is the weighting factor for radiation type R and $D_{T,R}$ is the absorbed dose for tissue T by radiation type R

Effective dose (Sieverts)

$$E = \sum_T W_T \cdot H_T \quad (7)$$

where, W_T is the weighting factor as given by ICRP 103 [37] and H_T is the equivalent dose for tissue or organ type T

One method of measuring organ dose in order to calculate effective dose is to use an anthropomorphic phantom for CBCT. This has been performed by a number of groups using small radiation dosimeters such as thermo-luminescent dosimeters (TLD) [15,22,23,31,36,39,40], fibre optic coupled water-equivalent plastic scintillators [41], silicon-photodiode dosimeters [42] and MOSFETs [43, 44].

An alternative method of estimating the effective dose is to use the ImPACT CT Patient Dose Calculator [45]. The ImPACT Patient Dose Calculator, designed originally for fan beam CT, uses a library of Monte Carlo calculated dose calculations [46] for organ doses in a humanoid mathematical phantom. The library covers numerous commercial CT scanners each characterised by the ratio of peripheral to central CTDI_w and central to in-air CTDI_w for both the head and body phantoms. To calculate the effective dose the operator selects the scanner type and then using a diagram of the humanoid phantom, the start and stop position of the scan required to include the organs covered in a typical scan. The software will provide individual organ doses with their weighting factors and equivalent doses as well as the total effective dose.

Ideally when using the ImPACT calculator the CT scanner for which the dose is to be calculated will be one of the scanners in the ImPACT library. If not, then the CT scanner can be matched to the closest one in the library using ImPACT factors derived from a linear combination of the ratios of the central and peripheral normalised CTDI₁₀₀ to CTDI_{air}. This method has been employed by several authors to match a CBCT scanner with fan beam CT scanners in the ImPACT library [22, 23]. Hyer *et al* compared the results of the ImPACT dose calculator with Monte Carlo measurements (described in 2.3. [47]). They measured in-air CTDI₁₀₀ and both central and peripheral CTDI₁₀₀ in the CTDI phantom to determine appropriate ImPACT factors for matching acquisition protocols, designed for the head, chest and pelvis with their respective acquisition settings e.g. fields of view on both the XVI and OBI CBCT systems, to a scanner in the ImPACT calculator. Organ doses from the ImPACT calculator were compared against previously published Monte Carlo calculated organ doses [41]. Hyer *et al* found that many organs agreed within 40% with generally better agreement for the pelvis scan

however, some discrepancies of more than 100% were found. These differences were attributed to: differences in the anthropomorphic phantom and the mathematical phantom; the use of the ImPACT calculator for calculation of CBCT dose when it is designed for fan beam CT dose calculation; the inability of the ImPACT calculator to cope with partial scans and finally, a probable calculation mistake identified in the calculation of the Thymus dose. They conclude that the ImPACT dose calculator is not suitable for calculating CBCT dose.

Gu *et al* have modelled both kV- and MV-CBCT systems using MCPNX [48] and applied these models to calculate organ doses to the VIP-Man phantom that was developed from the National Library of Medicine's Visible Human Project [49]. They concluded that the effective dose for the Head and Neck and prostate was 8.53 mSv and 6.25 mSv respectively for a 125kVp kV-CBCT exposure of 1350 mAs.

2.3. Calculation of patient specific dose from CBCT imaging

The above methods of calculating dose from CBCT are based on calculating the dose to a typical patient represented by either the head or body CTDI phantoms or the humanoid model of the ImPACT calculator. In the diagnostic world of radiation protection it may be sufficient to relate the dose for CT scan protocols used by a particular hospital to the radiation risk for the purpose of justification and for reporting dose. However, in radiotherapy where many CBCT scans might be performed during the course of treatment it may be necessary to calculate the dose to specific critical organs to ensure that the combined treatment and concomitant imaging dose does not exceed the dose criteria specified in the design of the treatment plan. In such cases individualised patient dose calculations may be required. There are two main research groups who have developed Monte Carlo (MC) models of CBCT imaging systems. Spezi *et al* have developed a model for the Elekta XVI CBCT scanner using EGSNRC/BEAMNRC and BEAMPP [8,50–52]. Ding *et al* have developed their MC model of the Varian OBI CBCT scanner using the EGSnrc/BEAMnrc code in their early work [53–55] and more recently with the BEAM/DOSXYZnrc code [56–58]. Chow *et al* have also modelled Elekta XVI using BEAMnrc and a Matlab based implementation of DOSXYZnrc [59].

Both Ding *et al* and Spezi *et al* have computed the dose on CT scans for a number of patients with a variety of anatomical locations and imaged using standard acquisition settings as supplied by the manufacturers [8, 60]. Both papers present doses to various organs of interest. A detailed analysis of their results is not presented here however, Table 2 summarises the dose for three anatomical sites. Note, the performance of the two CBCT systems should not be judged on the basis of the data reported in Table 2 since there is no reason to assume that the acquisition settings have been optimised to give the same trade-off between imaging dose and the quality of the image. This is evident in the head and neck doses where the dose for the XVI system is an order of magnitude less than the OBI system. Both authors present a number of interesting observations that are generally applicable to understanding patient dose from CBCT. Ding *et al* noted that the bone receives two to four times that of the soft tissue. Spezi *et al* showed that the bony structures consistently received a higher dose than the soft tissue with the mandible receiving three times the dose received by the surrounding soft tissues. This was caused by the increased mass-energy absorption coefficient due to the photoelectric interaction within the materials of higher atomic number. Spezi *et al* also showed that the addition of the bow-tie filter significantly reduced the dose by 22% in the Pelvis and 45% in the Chest. This was due primarily to the attenuation of the dose to the peripheral tissues but also the beam hardening affect. The bow-tie filter also reduces the scattered dose from the periphery of the patient to the imager which has the additional advantage of increasing image quality. Ding *et al* also calculated the dose for paediatric patients and observed that paediatric doses were of the order of two times that of an adult.

Table 2: Monte Carlo calculated patient doses (cGy) for three anatomical sites for the Elekta Synergy CBCT system and the Varian OBI CBCT system. Doses reported are for the body i.e. not to a specific organ.

	Pelvis/Abdomen	Head and Neck	Chest
Elekta XVI (Spezi <i>et al</i>) [8]	1.5 - 2.1	0.1 - 0.2	1.2 - 2.2
Varian OBI (Ding <i>et al</i>) [60]	1 - 5	3 - 9	2 - 9

Downes *et al* showed that the CBCT imaging dose has a left-right asymmetry due to the increased number of exposures at the start and stop gantry angle as the gantry rotation accelerates and decelerates at the beginning and end of each scan [52]. Unlike CT imaging where the patient is normally central in the CT scanner, in radiotherapy the isocentre is typically set to the centre of the target volume which may itself be offset from the centre of the patient's cross-section. Chow *et al* studied this effect and found for the pelvis phantom, variation in the mean dose of up to 20% for up to 10cm anterior-posterior shifts [61]. Dose variations for the chest and head and neck were typically between 7% and 17%.

One of the advantages of using Monte Carlo, apart from getting individualised dose measurements, is that dose distributions can be calculated for a range of situations including medium and large fields of view with offset collimation, partial arcs and offset isocentres.

An alternative to using Monte Carlo is to use a model based dose calculation similar to that used in planning the mega-voltage treatment. Aleai *et al* first showed that the then ADAC treatment planning system (TPS) , (now the Pinnacle TPS; Philips Medical Systems), can be adapted to calculate the dose at kilo-voltage energies by adjusting the photon spectrum and adding low dose scatter kernels to the collapsed cone superposition dose calculation algorithm [62-65]. This method has since been enhanced by Ding *et al* who added a medium dependent correction (MDC) to account for the increased photo-electric cross section of bone which leads to the increased bone dose previously observed with Monte Carlo effects [66]. The MDC algorithm also successfully accounts for upstream and downstream effects. The results show a significant improvement compared to the calculation using the collapsed cone superposition model alone and agreement with Monte Carlo measurements is good especially upstream and downstream of the bone.

2.4. Combining dose with treatment planning dose

For radiotherapy the risk of concomitant imaging needs to be considered in the context of the risk of secondary cancer induction from radiotherapy treatment. In addition the dose to critical organs already receiving high doses from the treatment needs to be assessed to ensure the additional imaging dose does not exceed dose limits. The imaging dose needs to be considered both within the treated volume and also peripheral to the volume.

Qiu *et al* calculated dose using Monte Carlo for relatively large volume gynae IMRT treatments with field length of ~15cm and for CBCT scans of length ~24cm) [67]. They concentrated on modelling the in-field dose, discussing out-of-field dose only briefly. In-field doses for organs at risk were calculated using Organ Equivalent Doses calculated using Linear, Linear-Exponential and Plateau radiobiological models. The greatest increment in dose from imaging one CBCT per fraction was 2.5% for the bowel with the linear model but this reduced to 1.3% for the plateau model. For dose in the peripheral region, the CBCT dose was compared with the linac scatter and leakage doses. In the peripheral low dose regions, where there is low risk of secondary malignancies, the incremental dose from CBCT was found to be an order of magnitude less than the IMRT scatter dose and less than or equal to the linac leakage dose.

Chow *et al* concentrated on in-field dose and compared CBCT dose with the treatment dose for a prostate IMRT case [59]. The PTV dose rose by 0.6 Gy (0.8%) for a 78Gy/39# treatment which

suggests the CBCT dose was ~ 1.5 cGy scan. The femoral heads saw the largest increase in dose of ~ 2.5 cGy (5%).

Perks *et al* measured the peripheral dose at the centre and on the surface of an anthropomorphic phantom [44]. They measured dose for a prostate IMRT treatment using MOSFETs and a kV-CBCT using TLDs. The dose from the IMRT dropped from the prescription dose of 2 Gy (per fraction) down to 1 cGy at 16 cm and 0.4 cGy at 21 cm distances from the field edge. In comparison, the CBCT was 0.5 cGy and 0.2 cGy at the same positions respectively (7 cm and 12 cm from the imaged volume). They used an S20 collimator which arguably provides a longer field of view than necessary for prostate IGRT. The nominal dose per scan was 6 cGy which they acknowledge was twice that normally used in their clinic. To put this into context 6 cGy is four times the UK's diagnostic reference level for imaging the abdomen/pelvis and is arguably 3-5 times higher than necessary for adequate image quality for CBCT image guidance [68].

Harrison *et al* published two papers on the subject of combined treatment and imaging doses covering anatomical sites of; larynx, breast and prostate [69, 70]. They compared imaging dose from 2D-portal imaging and 3D-CT imaging with the treatment dose. While this work was not based on CBCT the differences between CT and CBCT doses are likely to be minimal so the work provides a good perspective on the relative impact of kV-imaging on the combined treatment and imaging dose. For the prostate they measured dose using TLDs in the Rando phantom. Neutron doses were also calculated for the 15 MV beams. They calculated the dose to multiple organs both in-field and out-of field for combinations of a 37 fraction two phase prostate treatment with 26 CT images and 4 portal images. The excess relative risk was found to be < 0.1 for most organs with bone marrow showing the greatest increase in dose of up to 20%. They employed similar methods for the larynx and breast and concluded that the dose to critical organs increased by 5-20% with increases of up to 30% for bone surfaces and bone marrow. They noted that by far the larger component of dose to these organs was from scatter and leakage from the MV beam.

3. Clinical Consequences

3.1. Detrimental effects of radiation exposure

To date there has been no large scale epidemiologic studies of the cancer risks associated with CT scans. The evidence we have is derived from measurement and calculation of organ doses and applying organ-specific cancer incidence or mortality data derived from studies of atomic-bomb survivors [71]. The estimated attributable life time risk of death from cancer is $\sim 0.01\%$ increasing to 0.1% for exposures in early childhood [71]. However, these risks are calculated for the general population and not specifically for patients undergoing treatment with radiotherapy. Therefore the risk of imaging alone is small compared to the risk of treatment failure and other morbidities associated with radiotherapy treatment. It appears sensible that the risk from the imaging dose should be incorporated into the overall risk calculation including the treatment dose.

The primary risk of radiation exposure from radiotherapy, including any concomitant imaging, (excluding the risks of treatment failure and co-morbidities etc), is the induction of a secondary primary malignancy (SPM). As Tubiana notes in his review paper [11], these rarely occur before 10 years after treatment. However, with increased long-term survival rates the incidence of these malignancies is likely to increase. Tubiana found from cancer registries that the incidence could be as high as 20%. He also noted that SPM's tended to occur in tissues receiving more than 2 Gy. Data derived from the US SEER cancer registry by Berrington de Gonzalez *et al* found that 9% of 5 year survivors developed a solid tumour and that the relative risk was highest for tissues that typically received more than 5 Gy.

From the previous section we know that the imaging dose is typically small in comparison to the treatment dose. Nevertheless it adds to the radiation dose burden and contributes to the increased risk of SPM induction.

Very low doses are also associated with complications. Perks *et al* collated a number of such effects in their paper including: prolonged azoospermia at doses $> 2.5\text{Gy}$ [72]; loss of ovarian function at doses $< 2\text{Gy}$ [73] and hypothyroidism or thyroid nodules with median doses as low as 0.09 Sv [74]. Cataract formation can also occur with increasing likelihood for doses less than 2Gy [75].

3.2. Clinical Benefits.

As yet there is little published evidence on the clinical benefits of CBCT imaging. Chow *et al* calculated the NTCP increase from CBCT imaging during IMRT treatment of the prostate to be 0.5% although they recognised that the NTCP model used was relatively crude and did not take into account the RBE of kV imaging [59]. Nevertheless, they found that NTCP decreased by 3% when the CTV to PTV margin was reduced from 10mm to 5mm showing a net benefit of using CBCT imaging for every fraction of treatment. Kron *et al* showed that even when daily online IGRT (CBCT) was used with an adaptive strategy for bladder cancer, using plan libraries, integral dose to the whole irradiated volume and irradiated volume less the CTV, was less than the conventional treatment except in the case of some smaller volumes [76]. This is because on average the irradiated volume is smaller in the adaptive strategy than the irradiated volume that would be required to ensure the bladder is covered the majority of the time. Zelefsky *et al* compared cohorts of patients in which one group received prostate intensity modulated radiotherapy with IGRT and the other group received the same treatment but without IGRT [77]. They found that biochemical tumour control was significantly better for patients with high risk prostate carcinoma when IGRT was employed. In addition, late urinary toxicity was almost halved in the group with IGRT. While this study was performed using MV imaging and gold seed markers Moseley *et al* have demonstrated the equivalence of CBCT and gold seed marker/MV imaging based IGRT [78]. Recently, Bujold *et al* reviewed the literature and concluded that IGRT has enabled treatments such as hypo-fractionated stereotactic ablative radiotherapy of the lung, spine and liver [79]. They also conclude that,

“an improvement in relapse rate in prostate cancer, Hodgkin disease, and head and neck cancers using IGRT has been consistently reported.”

and that,

“There is a suggestion that prostate and head and neck cancer patients might have lower toxicity with IGRT, especially when combined with other technical advances like IMRT.”

4. Discussion

In this article various methods of calculating, estimating and measuring patient dose have been reviewed and the clinical consequences and benefits discussed briefly.

There is now a wealth of data measuring dose for the two main manufacturers of CBCT systems both using CDBI type techniques and using TLDs in anthropomorphic phantoms. Unfortunately, much of this data is difficult to relate back to protocols used on systems in the clinic as typically no reference dose is provided. Downes *et al* and Islam *et al* [52, 80] both suggest normalising doses to a reference dose measured in air. However, there is no standard employed by these authors. Such a standard would make interpretation of these results easier. In the absence of a standard reference dose, the results from each of these papers would have to be compared by normalising with the product of the exposure time and tube current.

Both IAEA and AAPM have produced recommendations on CTDI type measurements however there are foreseeable practical difficulties in implementing both and no published literature on the application of these methods to either the XVI or OBI CBCT systems. The AAPM-TG-111 report requires measuring the equilibrium dose from which the dose at any collimator setting can be determined. It is likely that this quantity, normalised to the dose equilibrium free in air, is transferable between machines of the same manufacturer/model. If so this could reduce the number of measurements required to provide dose calculations for all fields of view. The AAPM report discusses suitable phantoms but they are likely to be either excessively heavy due to their size or require filling

with water, neither of which is ideal. It favours a small Farmer chamber which suits the typically large cone angles (scan length) of these CBCT systems which is good in that nearly all radiotherapy centres will have a Farmer chamber but unless they also have a superficial X-ray unit they are unlikely to have suitable calibration factors to provide a dose traceable to a national standard. The IAEA approach has the advantage of requiring fewer measurements, by reducing the number required in a phantom and also only requiring a single CTDI phantom by choosing a short length reference collimation. However, on the Elekta Synergy CBCT system the shortest collimation for the small field of view is $\approx 13\text{cm}$ which does not fit within the 10cm chamber. Additional collimators would need to be manufactured to make this work. An additional complication is the use of the offset detector for medium and large fields of view. It could be argued that a suitably calibrated 10cm chamber is more readily available than a Farmer chamber but this is only likely to be in big radiotherapy centres with Medical Physics departments supporting Diagnostic Radiology as well as Radiotherapy. Smaller centres are likely to prefer the use of a Farmer chamber even if this requires additional calibration at low energies. However, the small chamber is not large enough or small enough to measure the CTDI for a narrow collimation required as a reference in the IAEA method.

To follow good practise such as defined in the UK IRMER legislation re optimisation of patient dose, imaging doses should be reported, image acquisition protocols should be optimised and the use of CBCT for IGRT should be justified on an individual patient basis. But what accuracy of dose measurement do we need? For reporting, maybe a CTDI type measurement would suffice or perhaps calculation of individual organ doses and effective dose for a population of patients using either TLD measurements in an anthropomorphic phantom or using a dose calculator such as the ImPACT is required or do we need to measure these doses for an individual patient using MC or a suitable kV dose model? On the one hand, the uncertainty in the current risk estimates for secondary primary malignancy induction are so large, that a CTDI type measurement would seem sufficiently accurate. Additional corrections for patient size could be used to refine the dose estimate. There would still be some residual uncertainties such as the variation of the dose due to the location of the isocentre within the patient and dose gradients due to partial arcs. On the other hand if the additional risks of concomitant imaging are to be fully understood maybe an accurate dose calculation including individual patient organ doses is required so that eventually the risks can be correlated, with sufficient specificity, to the delivered dose.

For quality assurance purposes and optimisation of imaging dose then a basic CTDI type dose calculation is probably sufficient. However, standardisation of the measurement is important if published techniques are to be compared. There is still a great deal of work required to both improve CBCT technology and its image quality capabilities to minimise the radiation dose required and understand the level of image quality required for image guidance which in some cases can be performed accurately by automatic image registration algorithms with much less dose than would be desirable for a human operator to perform manual image registration [12].

Finally, it is clear from the literature that there are identifiable risks attributable to radiation dose and that increased imaging is likely to increase those risks. However there still exists great uncertainty in estimates of risk for secondary malignancies from radiotherapy mainly due to the lack of epidemiological data and detailed understanding of the underlying mechanisms for cancer induction [81]. These uncertainties if converted back to dose are in many cases likely to be greater than the total concomitant imaging dose. This makes the process of justification difficult and inexact with the danger that those involved with prescribing radiotherapy and associated IGRT protocols become complacent and dismiss the imaging dose without due consideration. Extra care is required for justification of CBCT imaging of paediatric patients. However, in many cases, the likely benefits of performing IGRT using CBCT, including reduced CTV-PTV margins and reduced geographic miss, are likely to outweigh the detriments.

5. References

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