Imaging doses from the Elekta Synergy X-ray cone beam CT system

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Abstract

The Elekta Synergy is a radiotherapy treatment machine with integrated kilo-voltage (kV) X-ray imaging system capable of producing cone beam CT (CBCT) images of the patient in the treatment position. The aim of this study is to assess the additional imaging dose. Cone beam computed tomography dose index (CBDI) is introduced and measured inside standard CTDI phantoms for several sites (head: 100kV-38mAs, lung: 120kV-152mAs and pelvis: 130kV-456mAs). The measured weighted doses were compared to TLD measurements at various locations in a Rando phantom and at patients' surfaces. The measured CBDIs in-air at isocentre were 9.2, 7.3 and 5.3mGy/100mAs for 130, 120 and 100kV respectively. The body phantom weighted CBDI were 5.5 and 3.8mGy/100mAs for 130 and 120kV. The head phantom weighted CBDI was 4.3mGy/100mAs for 100kV. The weighted doses for the Christie hospital CBCT imaging techniques were 1.6, 6 and 22mGy for the head, lung and pelvis. The measured CBDIs were used to estimate the total effective dose for the Synergy system using the ImPACT CT Patient Dosimetry Calculator. Measured CBCT doses using the Christie hospital protocols are low for head and lung scans whether compared with electronic portal imaging (EPI), commonly used for treatment verification, or single and multiple slice CT. For the pelvis, doses are similar to EPI but higher than CT. Repeated use of CBCT for treatment verification is likely and hence the total patient dose needs to be carefully considered. It is important to consider further development of low dose CBCT techniques to keep additional doses as low as reasonably practicable.

1 Introduction

The Elekta Synergy^{TM} (Elekta Oncology Systems Ltd., Crawley, UK) is one of a new breed of radiotherapy linear accelerators specifically designed for image guided radiotherapy (IGRT). It has a kilo-voltage X-ray source and opposing amorphous silicon flat panel imager, both mounted at 90˚ to the treatment head for the acquisition of kV X-ray projection images for radiography and fluoroscopy, figure 1. The different interaction mechanisms of kilo-voltage (kV) photons with tissues and image transducers offer improved imaging compared to mega-voltage (MV) photons. This raises the prospect of enhanced localisation of target volumes and adjacent organs at risk, in the treatment room, compared to MV electronic portal imaging (EPI). Most importantly, the image sequence from rotation fluoroscopy can be used in filtered back-projection to reconstruct an X-ray volumetric image [1] . This form of cone beam computed tomography (CBCT) propels IGRT forward into a genuinely three dimensional (3D) technique and consequently has the potential for improving treatment setup and delivery [2-5].

In common with conventional trans-axial CT, the image quality for CBCT can be improved by increasing the number of X-ray projections acquired and the mAs used. However, the IRMER legislation [6] embodies the principle of patient doses being 'as low as reasonably practical' (ALARP). In practice this translates to justifying the risks to the patient in the context of the likely benefits from imaging protocols that should be fit for purpose, but no more than this. Precedent sets the starting levels for doses that might be acceptable in CBCT. CT image sequences for radiotherapy planning (RTP-CT) and an MV verification image involve absorbed doses to various critical structures of 1-40mGy and 10-20mGy respectively [7]. However, with CBCT in IGRT there is the potential to image a patient at every fraction in order to support set-up and the visualisation of the target/organs at risk (OAR) [3,8]. Guided solely by existing benchmarks the total additional dose resulting from serial use of CBCT could approach cautionary, deterministic levels. Since dose is added over a field of view wider than the target zone this has already influenced the debate on induced secondary cancers [7].

The Christie Hospital was one of the early sites selected to test the Synergy IGRT system which has been in clinical use since 2003. The aim of this paper is to report the likely patient doses arising from CBCT imaging protocols that have proven to be effective in clinical practice.

Figure 1: The Synergy system with retractable kV X-ray tube (right) and opposing kV imaging panel.

2 Theory

European Guidelines on quality criteria for conventional CT imaging have been published by the European Commission [9]. The guidelines describe a reference dose level termed the weighted computed tomography dose index ($CTDI_w$). This is a key component of the European dose reference levels (EDRL) for different anatomical sites. CTDI is simply the integral of the dose profile of a single slice along a line running parallel to the CT scanning axis, scaled by the nominal trans-axial slice thickness.

$$CTDI = \frac{1}{L} \int_{-Z/2}^{Z/2} D(z)\, dz \quad [mGy] \qquad (1)$$

where D(z) is the dose profile and Z is the integration range that covers most of the penumbra. The International Electrotechnical Commission (IEC) recommends Z=10cm. The Dose profile integral (DPI) is usually measured using a 10cm long ionization chamber in CTDI. L represents the nominal slice thickness (or the total thickness of more than one slice for multi-slice CT). For CBCT, L should represent the nominal length of the FOV in the axial direction and Z should be long enough to cover the penumbra region. This can be up to 40cm for the Synergy system. Clearly, given the scale and that CBCT is not a sequential, slice based technique, CTDI is impractical for measuring dose in CBCT. Instead we suggest the continued use of the standard 10cm chamber for CBCT dose measurement, whilst acknowledging that this method will not account for the dose in the penumbral region. Nevertheless, it will provide a reasonable measure of the dose in the central 10cm region of the FOV. To distinguish this dose from CTDI, we will refer to it as cone-beam dose index (CBDI). In this case, L in equation (1) represents the chamber length of 10cm.

Similar to $CTDI_w$, $CBDI_w$ reflects the variation of dose deposition at depth by differentially weighting peripheral (p) and central (c) doses measured in standard CTDI phantoms.

$$CBDI_w = \frac{1}{3} CBDI_c + \frac{2}{3} CBDI_p \quad [mGy] \qquad (2)$$

The $CBDI_w$ is normalized to 100mAs by dividing the $CBDI_w$ by the exposure E (mAs) used to measure CBDI

$$_n CBDI_w = \frac{CBDI_w \times 100}{E} \quad [mGy/100mAs] \qquad (3)$$

3 Materials and Methods

3.1 Synergy kV X-ray IGRT system

The Elekta Synergy system, release 3.1, used in this study has a Dunlee X-ray tube (DU 304, Dunlee, USA) with a focal spot size of $1.2mm^2$, located 100cm from the centre of CBCT rotation. The tube has 1.5mm Al equivalent inherent filtration and additional compound filtration of 1.5mm Al and 0.127mm Cu. Tube potentials range from 40-130kVp. Exposures are pulsed and range from 0.1-3.2mAs per X-ray projection. Two pairs of symmetric collimators (standard diaphragms) can be adjusted manually to shape the beam. The imaging transducer is of the

indirect kind with a Gd_2O_2S:Tb amorphous silicon (AmSi) flat panel (RID 1604, Perkin-Elmer Optoelectronics, Wiesbaden, Germany). It has an active area of $41x41cm^2$ addressed as an array of 1024x1024 pixels, each pixel having 400μm pitch. The panel is located 53.6cm from the axis of rotation and images are captured at a fixed frame rate of 2.7Hz. Production Synergies are currently being installed with a CsI panel operating at higher frame rates. The latter will be the subject of a future, comparative investigation.

3.2   Dose measurements

At the outset it was regarded as important to attempt to identify the characteristic of skin and central doses that would assist practical prediction of patient related CBDI and the subsequent selection of the most efficient technique settings for clinical deployment. For the initial investigations, technique settings were: 100kV, 94mAs (25mA, 10ms/projection, nominal 380 projections) for the head; 130kV, 199mAs (40mA, 13ms/projection, nominal 380 projections) for the lung; and 130kV, 456mAs (40mA and 30ms/projection, nominal 380 projections) for the pelvis. Clinical investigations subsequently used the same settings for the pelvis but different settings (100kV, 38mAs) for the head and (120kV and 152mAs) for the lung. With IRMER in mind, these settings were found to be sufficient for IGRT at these sites. Measurements were taken to assess patient dose from the Synergy CBCT system using a Rando anthropomorphic phantom, standard CTDI phantom and patients. A 26cm-diameter and 26cm-long FOV in the axial direction was used for all dose measurements unless stated otherwise.

3.2.1   Rando Phantom

Lithium fluoride thermo-luminescent dosemeters (Harshaw TLD-100 LiF) were placed on the surface and at various depths inside a Rando phantom. Surface and internal doses to the head, lung and pelvis were investigated. The TLDs were read with a Harshaw 4500 reader (Harshaw Thermo Electron, Solon,USA). The TLDs were calibrated using a diagnostic X-ray set with 3mm Al filtration. The X-ray beam air kerma was measured with a 6cc chamber (Radcal model 90X6-6) and a Radcal 9010 radiation monitor. The instrument calibration is traceable to National Standards. The overall uncertainty in the TLD reading is ±10% at 10mGy at the 95% confidence limit. Figure 2 shows a picture of the Rando phantom and the positions at which doses were measured for CBCT scanning. Each scan consisted of 380 projections taken during a 360˚ gantry rotation.
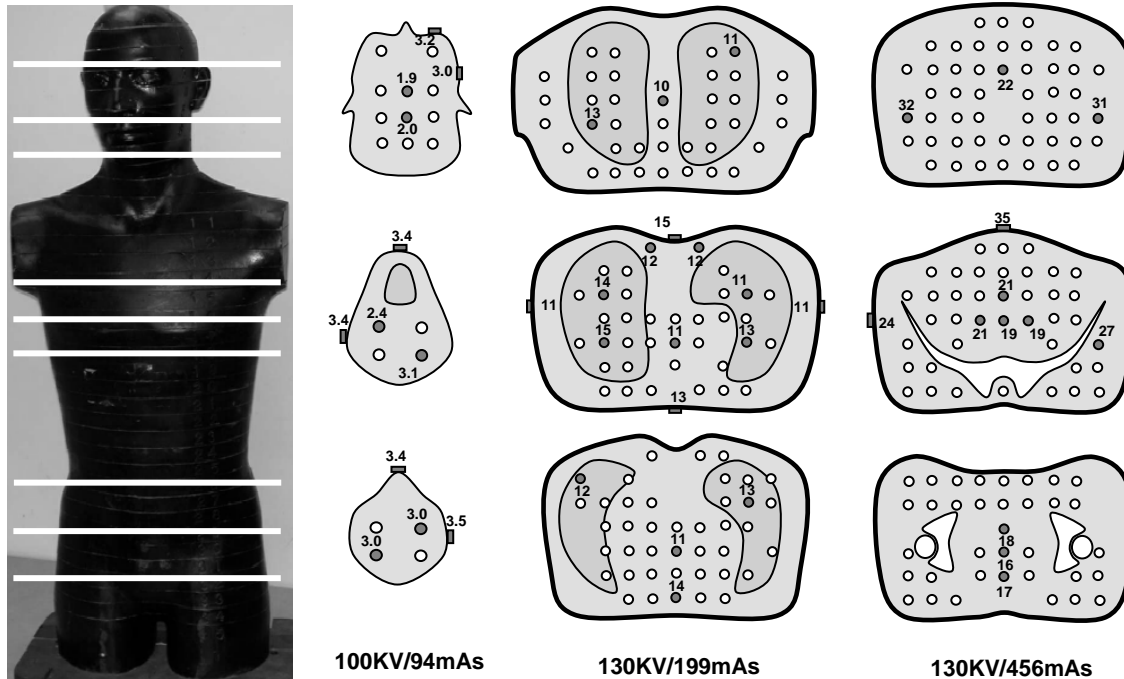
Figure 2: TLD measured dose [mGy] in the Rando phantom. White lines indicate the slices where doses were measured. The locations within each slice are indicated as black points, with the dose measurement itself adjacent. The kV and total mAs for each scan are indicated.

3.2.2   *CTDI phantom*

CBDIs were measured for a range of kV, mA and ms settings in air and then inside standard, 14cm long, 16 and 32cm diameter head and body CTDI phantoms. In contrast to conventional slice-based CT scanners, which irradiate only short lengths of the patient at any given instant, the use of the standard 14cm long CTDI phantoms does not cover the wide field CBCT and underestimates the CBDI due to missing scatter [10]. Hence, the effective length of the CTDI phantom was increased by adding 15cm of Perspex (Plexiglass) to both ends of the body phantom and one end of the head phantom. A dose meter (Radcal 9010, Radcal Corp, Monrovia, CA, USA) with a 10cm long, 3cc ion chamber was used. For the CBDI measured in air ($CBDI_{air}$), the ion chamber was placed at the centre of the CBCT rotation plane using the treatment machine alignment lasers, and with the chamber axis parallel to the axis of rotation. In the case of CBDI measured in a phantom, the centre of the phantom was positioned at the centre of rotation, again with axis parallel to that for rotation. Doses were then measured at the centre and periphery of the phantom for the standard 26cm long FOV and also for collimated fields of view in the axial direction of 21 and 16cm.

A $0.125cm^3$ ionisation chamber (PTW, Freiburg, Germany) was used to measure the central and peripheral dose profiles across the CTDI body phantom. This enables investigation of the shape of the dose profile, calculation of $CTDI_w$ and comparison with $CBDI_w$. The adequacy of using a 10cm long ionisation chamber in CBCT dose measurements was assessed.

3.2.3 Patient Surface

Point doses from CBCT scanning at patients' surfaces were measured using TLDs. Because of the kV energy of the CBCT X-ray beam, the dose at the surface represents the maximum dose that the patient receives. TLD measurements were taken for a total of 9 patients treated for head and neck, lung, bladder, prostate or cervix cancer with three to five TLDs per patient. These patients were from a pilot study looking at the use of cone beam 3-dimensional imaging in radiotherapy. The study had Local Research and Ethics Committee approval (South Manchester Local Research Ethics Committee approval number 02/SM/375). Informed written consent was obtained from all patients.

3.3 ImPACT Dose Estimation

The ImPACT CT Patient Dosimetry Calculator (ImPACT, London, UK) uses Monte Carlo generated dose data (NRPB SR-250 [11]) for a geometrical human model, and $CTDI_{air}$ values for a particular scanning system, to estimate the absorbed dose to different organs and to calculate the total effective dose. The software package (version 0.99v) [12-14] was used to calculate these doses from the Synergy system. The doses are approximate, since the ImPACT calculator and the NRPB datasets do not take into account the divergent nature of a cone beam and so the shape of the dose distribution at the ends of the field is not accurately modelled. The Synergy system was matched to the nearest equivalent CT scanner within ImPACT database. The ratios of the CTDI, at centre and periphery, to air measurements allow the calculation of an empirically derived 'ImPACT factor' according to the formula available in the ImPACT calculator. The scanner with the closest factor having a flat beam filter as listed in NRPB SR-250, was selected.

4 Results

4.1 CBDI measurements

Figure 3 shows a longitudinal relative dose profile through more than half of a 26cm long imaging field, measured using a 0.125cc chamber at central and peripheral positions, in a 32cm diameter CTDI phantom with 15cm Perspex scatter material added to both ends. The reduction in the dose profile at the periphery (inside the FOV) and the start of the sharp decline of the profile at 11cm from the FOV centre, rather than 13cm the nominal width, are attributable to the reduced scatter near the edge of the FOV and the divergent beam in CBCT. If one considers a dose profile from a full rotation scan, the dose profile at the periphery is influenced far more by the narrower entrance beam than the wider, highly attenuated exit beam. This is why the dose profile at the periphery is smaller than the nominal width for the field of view. By definition, the weighted dose is influenced more by the peripheral dose and hence the shape of the weighted dose profile is closer to the peripheral dose profile. Assuming symmetry of the relative dose profile, the weighted dose ($CBDI_w$) across the central 10cm of the FOV (the length of a standard ionisation chamber usually used in CTDI measurements) is 9% higher than the weighted dose ($CTDI_w$) calculated from the integration of the weighted dose profile across 40cm and divided by 26cm (the nominal length of the imaging field). It can be seen from the relatively flat weighted-dose profile of figure 3 that the use of a 10cm chamber gives a useful, conservative overestimate of the weighted dose across the whole field of view (FOV).
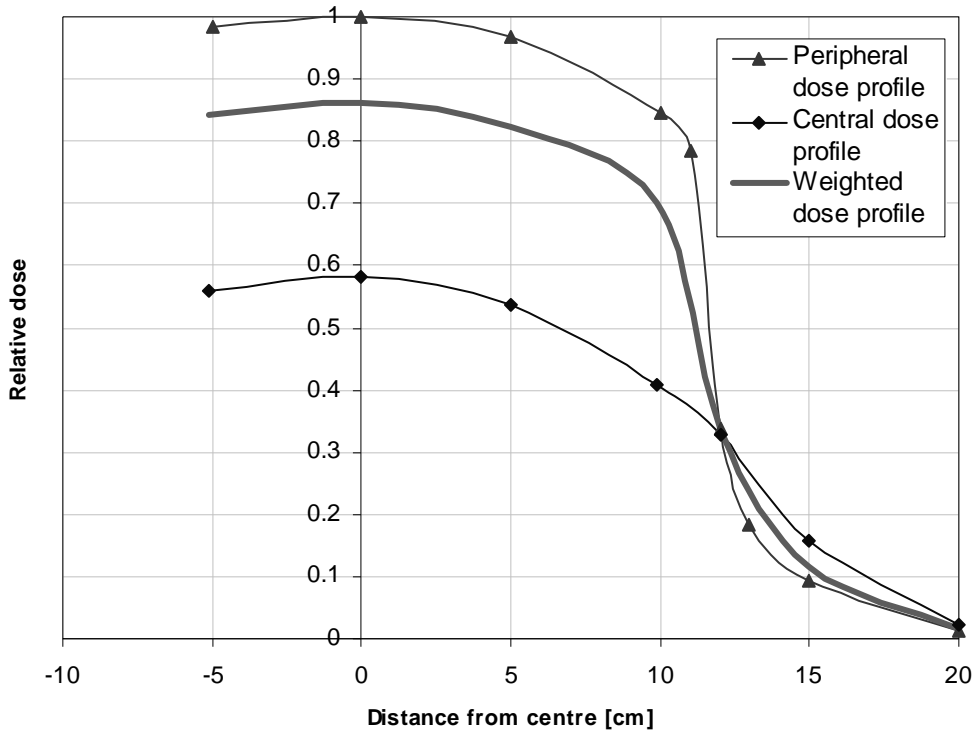
Figure 3: Longitudinal dose profiles across a 26 cm imaging field normalised to the maximum measured dose. The dose profiles were measured at central and peripheral positions in 32cm diameter CTDI phantom with 15 cm additional scatter material to both ends of the phantom.

Table 1 shows the $_nCBDI_w$ values measured for the beam settings adopted at the Christie Hospital. The $CBDI_w$ for 26cm long FOV scans are 1.6, 6 and 25mGy for head, lung and pelvis respectively. The uncertainty in the CBDI measurements is ±3% (1 standard deviation) estimated from repeated measurements. CBDI measurements obtained for the CTDI phantom without additional scatter material were lower, by 31% and 8% at the centre and periphery respectively, compared to those measured using additional scattering material on both sides. Beam collimation in the longitudinal direction reduces the contribution of scatter to CBDI. Compared to the 26cm long FOV, 21 and 16cm long FOVs resulted in 6 and 12% reduction in $_nCBDI_w$ respectively

Table 1: Beam settings, nCBDIw values and weighted dose estimates for the clinical CBCT scanning protocols adopted at the Christie Hospital.

Site	Beam settings		nCBDI_air	nCBDI_w	CBDI_w
	kV	mAs	(mGy/100mAs)	(mGy/100mAs)	(mGy)
Head	100	38	5.3	4.3 (head)	1.6
Lung	120	152	7.3	3.8 (body)	6
Pelvis	130	456	9.2	5.5 (body)	25

4.2 Rando and Patient doses

Figure 2 shows the TLD dosimetry results at different locations in the Rando phantom. The average doses from CBCT at the anterior and lateral surface of several cancer patients are provided in Table 2. The Rando and patients' measured point doses were normalised to the adopted imaging beam settings ($Dose_{normalised}=Dose_{measured}\times(mAs/mAs_{adopted})\times(kV/kV_{adopted})^{2}$) for comparison, and are given in Table 3. There is very good agreement between the external doses measured at the surface for Rando and patients. Hence, the internal Rando point doses represent a good estimate of the patient internal point doses.

Table 2: Summary of the TLD measured dose in mGy at the surface of patients imaged with CBCT.

Patient	Site	TLDs / patient	mAs	Dose at surface (mGy)	
				Anterior	Lateral
1	Prostate	3	440	32	21
2	Prostate	3	440	34	22
3	Bladder	5	440	33	21
4	Prostate	2	440	35	23
5	Cervix	4	440	35	21
6	H&N	3	90	3.0	2.9
7	H&N	2	90	2.7	2.8
8	Lung	2	200	13	7.6
9	Lung	3	200	15	8.1

Table 3: Comparison of the Patients, Rando and ImPACT doses in mGy at different locations normalised to the beam settings of the adopted imaging protocols.

Site (beam settings)	Position	Patient	Rando	ImPACT
		mGy	mGy	mGy
Head	Surface (Ant)	1.2	1.3	-
(100kV/38mAs)	Surface (Lat)	1.2	1.2	-
	Eye lens	-	1.3	1.2
Lung	Surface (Ant)	11	10	-
(120kV/152mAs)	Surface (Lat)	6.0	7.2	-
	Heart	-	7.2	7.1
Pelvis	Surface (Ant)	35	34	-
(130kV/456mAs)	Surface (Lat)	22	23	-
	Uterus	-	21	23

4.3 ImPACT dose estimation

Based on the CBDI measurements, Synergy was matched to the Siemens Somatom, DRH, CR conventional scanner. Using the adopted imaging beam settings, the ImPACT calculator estimated doses to be 1.2, 7.1 and 23mGy for eye lenses, heart and uterus. These are in good agreement with 1.3, 7.8 and 21mGy measured using TLDs in Rando after normalisation to the adopted technique settings. The total effective dose calculated using the ImPACT software were 0.1, 1.6 and 6mSv for the head, lung and pelvis scans respectively.

5   Discussion

The choice of the nominal FOV width for CBCT beam is not a clear cut because of the divergence of the conical beam. We feel that the width of the beam at the axis of rotation gives a balanced choice in terms of dose assessment and available useful image information. Figure 4 (a) shows the entrance, isocentre and exit beam widths arising from rotational scanning with a conical beam. There are always a full set of projections for points on the conical apices defining the scanned volume. Additionally, there are sufficient projections from most of the scanning directions for a good if not perfect reconstruction of adjacent points. When viewed non-axially, a 26cm field of view clearly provides a substantial amount of useful information, well beyond the entrance width. This is demonstrated in figure 4 (b) which is a coronal 256×256mm CBCT image. On this image the edges of the divergent X-ray beam used for cone beam scanning are indicated. It is of course possible to exclude areas not reconstructed from a complete set of projection angles, but we feel this is short sighted, since it discards useful and available data!
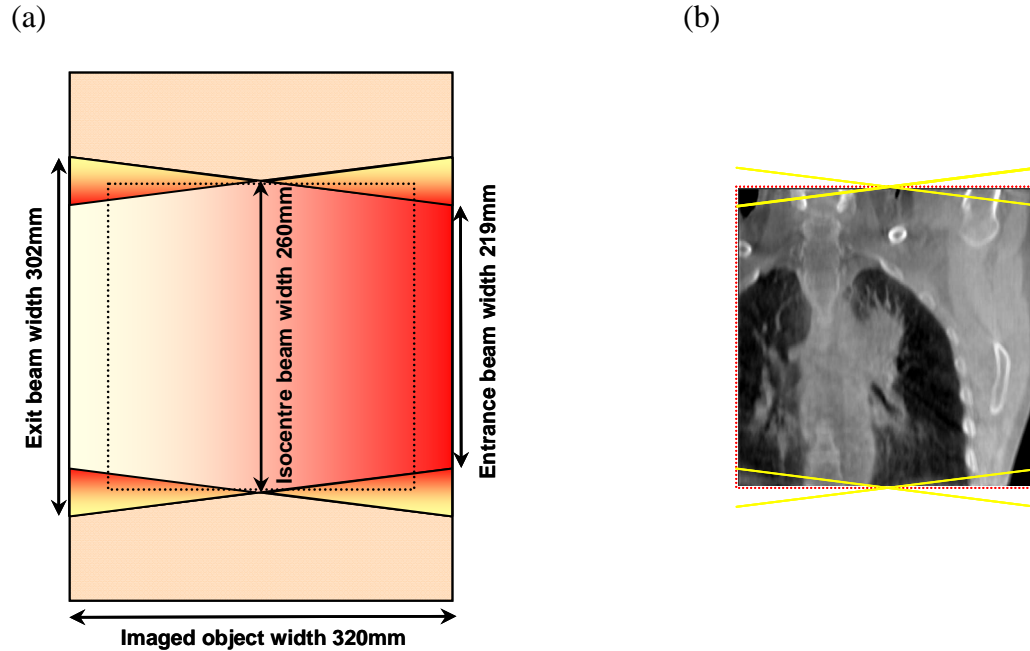
(a)

(b)

Figure 4: (a) A schematic diagram of the entrance, isocentre and exit beam width when imaging a 32cm wide object. (b) A coronal 256×256mm image of a lung cancer patient showing useful information beyond the nominal entrance beam width.

In the case of CTDI dose measurements for diagnostic CBCT, Mori, et. al. [10] pointed out the need for a longer CTDI phantom and ionisation chamber. Our results from the Synergy system strongly reinforce the need for a longer CTDI phantom (or the addition of scattering material) in radiotherapy CBCT, where the image volume is generated from a single rotation of a genuinely wide-angle cone beam. However, from the measured shape of the weighted dose profile across the CTDI phantom, figure 3, the use of a 10cm chamber appears to give a conservative overestimate measure of the weighted dose across the imaged volume.

Currently, patient position verification is commonly done by acquiring EPI using the megavoltage X-ray treatment beam. These images are acquired using the exit treatment beam but in many cases the limited field provides insufficient detail to register to the digitally reconstructed radiograph (DRR). Hence, EPIs with wider fields are often taken before treatment. At our institution, linear accelerators are calibrated to produce a point dose of 1cGy per monitor unit at the depth of maximum dose for a 10cm×10cm field and a 100cm focal to surface distance. EPIs are usually acquired with 2-4 monitor units per image and often two orthogonal EPIs are needed to verify the patient position in the three cardinal directions. Approximating the average patient dose to half of the maximum, then the average dose from verification images is typically 20-40mGy per pair of images. The CBDI dose from CBCT is less than that from wide-field EPI, especially for head and lung scans. This advantage, in addition to the better image quality using kV X-rays and the 3D information it provides, suggest it may become an important modality for verification in the future.

The European directive [15] and UK regulations [6], require the justification and optimisation of dose used in medical imaging. The justification for the extra dose from CBCT could be the

reduction of uncertainty in the process of treatment delivery which promises higher rates of tumour control and/or lower rates of complication. Nevertheless, the CBCT image quality should always be optimised for a given purpose. For example, verification of head and neck position can be achieved with a lower CBCT dose (~1mGy), sufficient to show bony details to register with the planning scan [16], than that required for the visualisation of tumours or soft tissue structures in the pelvis (~25mGy). Other methods should be sought to keep the dose as low as practically achievable.

More consideration is required when contemplating repeated or serial use of CBCT. For example, a typical CBCT imaging protocol for pelvis would result in patient surface dose of 30mGy per scan [8,17]. In the case of an online correction protocol or clinical studies that require imaging on a daily basis (40 fractions), the total surface dose approaches the deterministic level for transient skin erythema of 2Gy [18]. Caution should be exercised when imaging sites such as breast, particularly in younger women, where the skin dose from radiotherapy alone is already of clinical concern [19]. For head and neck, ocular opacities have a threshold of only 500mGy [18]

Figure 5 shows image quality for CBCT imaging protocols adopted at the Christie Hospital. The CBDI from these protocols are very low for the head and lung scans compared with the CTDI of two recent surveys conducted in the UK and Europe for single and multi-slice CT [20, 21]. The CBDI from pelvic scans is higher due to the need to see soft tissue contrast in what is after all a high scatter imaging modality. Table 4 gives a comparison of the CBDI with the CTDI of the two surveys.

Figure 5: Representative axial slices through CBCT image volumes of head (left), lung (middle) and prostate (right) showing the image quality with the Christie Hospital imaging protocols. Effective slice widths are 3mm for the head and 5mm for lung and pelvis.

Table 4: Comparison of the cone beam dose index (CBDI) in mGy with UK and European surveys of volume computed tomography dose index (CTDI$_{vol}$ = CTDI$_w$/pitch, where pitch is the ratio of the table feed and slice thickness).

	Europe MSCT CTDI$_{vol}$ [mGy]	*UK* SSCT CTDI$_{vol}$ [mGy]	*UK* MSCT	*Christie* CBCT CBDI [mGy]
Head	77	59	80	**1.6**
Lung	12	10	13	**6**
Pelvis	16	13	14	**22[a]**

Abreviations: Single slice CT (SSCT), multi slice CT (MSCT), cone beam CT (CBCT).
[a] Dose from a collimated FOV in the longitudinal direction (16cm) used clinically.

6 Conclusion

Measurements of CBCT dose made in Rando, CTDI phantom and patients were generally consistent with each other. For image acquisition settings adopted at the Christie Hospital, CBCT doses from the Synergy image guided radiotherapy system are low for sites such as head and lung, by factors >10 and 2 respectively, compared to conventional CT. For sites where soft tissue details are needed, such as the pelvis, doses are ~50% higher than those seen in conventional CT and similar to the dose from orthogonal EPIs minimally required for 3D setup verification.

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