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Estimation of effective imaging dose for kilovoltage intratreatment monitoring of the prostate position during cancer radiotherapy

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Abstract

Kilovoltage Intratreatment Monitoring (KIM) is a novel real-time localization modality where the tumor position is continuously measured during Intensity Modulated Radiation Therapy (IMRT) or Intensity Modulated Arc Therapy (IMAT) by a kilovoltage (kV) x-ray imager. Adding kV imaging during therapy adds radiation dose. The additional effective dose is quantified for prostate radiotherapy and compared to dose from other localization modalities. The software PCXMC 2.0 was used to calculate the effective dose delivered to a phantom as a function of imager angle and field size for a Varian On-Board Imager. The average angular effective dose was calculated for a field size of 6 cm × 6 cm. The average angular effective dose was used in calculations for different treatment scenarios. Treatment scenarios considered were treatment type and fractionation. For all treatment scenarios, (i.e. conventionally fractionated and SBRT; IMRT and IMAT), the total KIM dose at 1 Hz ranged from 2–10 mSv. This imaging dose is less than the Navotek radioactive implant dose (64 mSv) and a standard SBRT CBCT pretreatment scan dose (22 mSv) over an entire treatment regime. KIM delivers an acceptably low effective dose for daily use as a real-time image-guidance method for prostate radiotherapy.

Keywords

effective dose; gantry-mounted kV x-ray imager; intratreatment motion; real-time tumour localization; Monte Carlo simulations

Introduction

There are a number of modalities available or under development for the real-time measurement of prostate positions during radiotherapy treatments. These include MV imaging (Herman *et al.*, 2001), combined kV/MV imaging (Cho *et al.*, 2009), ultrasound (Schlosser *et al.*, 2012), magnetic resonance imaging (MRI), implanted radioactive fiducials

(e.g. Navotek (Shchory *et al.*, 2010)) and electromagnetic guidance (e.g. Calypso (Kupelian *et al.*, 2007)). Several of these modalities require the implantation of one or more markers into the prostate. The real-time localization modality studied in this work involves a gantry-mounted kV imager acquiring regular x-ray projections of prostate markers and reconstructing the 3D position based on the 2D projections. This modality is termed Kilovoltage (kV) Intratreatment Monitoring (KIM) throughout this manuscript (Poulsen *et al.*, 2008; Poulsen *et al.*, 2009; Poulsen *et al.*, 2010a; Poulsen *et al.*, 2010b). KIM is a novel real-time tumor localization modality in which kV images are acquired during treatment i.e. when the MV treatment beam is on. The imaging frequency can be specified and is typically between 1 to 10 Hz.

The advantages of this modality are:

- Accuracy: Sub- mm accuracy can be achieved in-phantom and in-patient for prostate radiotherapy (Ng *et al.*, 2012; Poulsen *et al.*, 2010b)
- Widespread applicability: this modality is achievable on existing linear accelerators (linacs) with gantry mounted x-ray imaging panels, which includes the majority of new accelerators being installed.
- Low barrier to clinical implementation: To operate KIM only involves the addition of image processing software without any hardware modifications.

However, an important consideration of the KIM modality is the additional radiation dose delivered to the patient. Imaging dose is gaining broader public attention. In addition to the 'Image Gently' and 'Image Wisely' campaigns, the US Food and Drug Administration (FDA) recently announced their 'Initiative to Reduce Unnecessary Radiation Exposure from Medical Imaging'. Therefore the quantification of imaging dose, and ultimately the means to reduce this dose, is of high public significance.

Although AAPM TG-75 has previously quantified imaging dose due to kV Cone Beam Computed Tomography (CBCT) (Murphy *et al.*, 2007), calculations of effective dose are limited. Table 5 in the Appendix shows that a variety of studies on CBCT dose have determined the kV dose either via experiment or Monte Carlo (MC) simulations. Whilst most of these studies are indeed quite detailed, it is noteworthy that a number of them that were included in TG75 (e.g. Islam *et al.* (Islam *et al.*, 2006), Amer *et al.* (Amer *et al.*, 2005) and Cossmann *et al.* (Cossmann *et al.*, 2005)) did not determine effective dose, only absorbed dose. This necessitates the use of the table of conversion coefficients of dose-area product calculated by MC simulations in 1992 by Le Heron *et al.* (Heron, 1992), who only computed the coefficients for AP, PA and lateral projections. This is insufficient for the determination of KIM dose as the majority of beam angles (e.g. during IMRT) are neither AP, PA nor lateral. For the studies that determined effective dose, Endo *et al.* (Endo *et al.*, 1999) estimated effective dose, but only for a prototype Computed Tomography (CT) unit and dose to the pelvic region was not determined. Hyer *et al.* (Hyer *et al.*, 2010) estimated effective dose to the pelvis, but did not study the dose dependence on projection angle. Furthermore, their field size was that of a conventional pelvic CBCT scan, i.e. much larger than that required for KIM guidance of the prostate (6 cm × 6 cm). Gu *et al.* (Gu *et al.*, 2008) simulated the effective dose delivered for a prostate scan. However, the field size used (17

cm $\times$ 25 cm) was much larger than that required for KIM. He *et al.* (He *et al.*, 2010) simulated effective dose using PCXMC as a function of different projection angles. However they did not assess dose to the pelvic region. More recent MC studies (e.g. Spezi *et al.* (Spezi *et al.*, 2012), Ding *et al.* (Ding and Coffey, 2009; Ding *et al.*, 2008) and Chow *et al.* (Chow *et al.*, 2008)) while detailed, do not provide the effective dose and are hence insufficient for quantifying imaging dose during KIM. Crocker *et al.* (Crocker *et al.*, 2012) measured the KIM absorbed dose for a cohort of prostate cases. In particular, the variation of absorbed dose with a) population minimum field sizes, b) patient minimum field sizes and c) beam minimum field sizes was measured. The absorbed dose as a function of treatment duration was also determined. However, no estimation of effective dose was conducted in that study. There is thus a dearth of data for the effective dose from small field x-ray beams directed at the prostate with varying imaging angle.

Therefore, the goal of this study was to quantify the additional imaging effective dose to the patient using KIM as a function of imager angle, field size, treatment type, fractionation, treatment duration and imaging frequency.

Methods and materials

2.1. X-ray beam and treatment parameters

In this study, the characteristics of the Varian OBI (On Board Imager) were used in MC simulations, described in II.B. The Varian OBI is a kV x-ray imager mounted on the linac gantry. The parameters considered were divided into two categories, namely x-ray beam and treatment parameters. The x-ray beam parameters are the filtration, anode angle, tube voltage, tube current, exposure time, imager gantry angle, source to isocenter distance (SID), source to detector distance (SDD) the field size at isocenter and the number of photons in the MC simulation. The treatment parameters are the treatment type, fractionation, treatment duration and imaging frequency. These parameters are shown in table 1.

The filtration (no bowtie included) and anode angle were specific to the Varian OBI x-ray tube characteristics. The tube voltage, current and exposure time used in the simulations were based on the Varian CBCT pelvic scan protocol. This protocol was used because preliminary clinical results suggest that these settings are suitable for marker detection. The kV source angle was varied from 0°–360° in steps of 10°. The SID was set to 100 cm and the SDD to 180 cm. The isocentre was set to the geometric centre of the prostate.

The field size at isocentre was varied from 4 cm $\times$ 4 cm to 8 cm $\times$ 8 cm in 0.5 cm $\times$ 0.5 cm increments, giving a total of 9 field sizes. These were based on the quantification of the field sizes needed to image the implanted markers through a review of 22 prostate patient CT images (Crocker *et al.*, 2012). Prostate images with three implanted markers were analysed and intermarker separations for each patient were measured. The largest field size with an added motion margin of the 22 patient dataset was ~6 cm $\times$ 6 cm (Crocker *et al.*, 2012). The number of photons was set to 3×10^6 per projection to give a statistical uncertainty of ~1%. This uncertainty was small compared to changes in dose resulting from varying the other parameters in this study.

The average angular effective dose per projection was obtained from the MC simulation results. To translate this into patient treatments, we simulated several clinical scenarios by varying the treatment parameters. The treatment parameters varied were treatment type (IMRT or IMAT) and fractionation (conventional or Stereotactic Body Radiotherapy (SBRT)).

The number of fractions for conventional fractionation is commonly 39 and for SBRT it is typically 5 (Friedland *et al.*, 2009; Katz *et al.*, 2010; King *et al.*, 2009; King *et al.*, 2011; Madsen *et al.*, 2007). For the treatment durations, the measurements of Oliver *et al.* (Oliver *et al.*, 2009) were used. For conventional fractionation, the average total treatment durations were determined from multiplying the duration per fraction and the number of fractions. These were:

- 186.2 minutes for IMRT (average of 5 and 9-field) and
- 84.8 minutes for IMAT (average of single and double arc).

These values were obtained for a dose rate of 400 MU/min. For SBRT, a conservative upper estimate of treatment delivery is 3.25 times the conventional fraction delivery time (6.5Gy/fx vs 2Gy/fx) i.e. it is a simple linear increase based on the prescription dose. Hence, the average treatment durations for SBRT were estimated to be:

- 77.6 minutes for IMRT
- 35.3 minutes for IMAT

The imaging frequency was determined by considering simulations performed by Poulsen *et al.* (Poulsen *et al.*, 2009). Here, the imaging frequency is defined as the number of kV images acquired in 1 second. In this study, a rotating imager was used to track the simulated trajectory of a prostate. The position estimation accuracy was investigated as a function of imaging frequency. The accuracy was quantified by the mean 3D root-mean-square error (RMSE). It was found that the mean 3D RMSEs for 0.5 Hz, 1 Hz and 5 Hz were 0.26 mm, 0.23 mm and 0.22 mm, respectively. Based on these results, sufficient accuracy (0.23 mm RMSE) is achieved with 1 Hz imaging, and therefore a higher imaging frequency would add additional dose without a commensurate increase in treatment accuracy. Given the high accuracy of the 0.5 Hz and 1 Hz methods, it is prudent to consider lower imaging frequency strategies to reduce the patient imaging dose. Given the low (0.26 mm RMS) error at 0.5 Hz, the frequency of 0.1 Hz was included in this study. However as yet, there are no estimates of the loss of treatment accuracy for imaging frequencies lower than 0.5 Hz.

2.2. PCXMC and phantom

The Monte Carlo program PCXMC 2.0.1.3 is a software package that can be used to calculate averaged organ absorbed doses and effective doses to a patient undergoing diagnostic x-ray kV scans. The software is capable of simulating an x-ray beam irradiating a mathematical human phantom and calculating the absorbed and effective dose (as defined in the International Commission on Radiological Protection (ICRP) Publication 60 (ICRP, 1991)) deposited in each organ and in the whole body. The tissue weighting factors used for the effective dose calculation were recommended by ICRP Publication 103 (ICRP, 2007).

The mathematical human phantom employed in PCXMC was developed by Cristy and Eckerman (Cristy and Eckerman, 1987). The longitudinal cross section of the phantom is an ellipse. The dimensions and mass of this phantom were standardised and are shown in table 2. This standard adult phantom was used in all our simulations. The shape, size and location of the prostate with respect to other organs was modelled based on CT and MRI images (Moller and Reif, 1994b).

PCXMC allows for the patient irradiation geometry to be defined. In particular, the isocenter was specified to be at the geometric center of the prostate. The field sizes at the isocenter can also be varied. Specifying these parameters permits simulation of the irradiation geometries of the Varian OBI.

PCXMC only considers photoelectric, Rayleigh and Compton scattering interactions. Other interactions are not considered as the maximum energy is limited to 150 keV. At these energies, the range of secondary electrons is only a fraction of a millimetre so their energies are assumed to be deposited at the site of the photon interaction. At each interaction point, the energy deposited to the organ at that point is calculated and stored for dose calculation. A large number of independent random photon histories are generated. The mean energy deposited in each organ is used to calculate the dose to those organs. No specific data packages are used in PCXMC.

PCXMC was developed by the Radiation and Nuclear Safety Authority of Finland (STUK) and has been used in previous imaging dose studies (Helmrot et al., 2007; Schultz et al., 2003; Schmidt et al., 2000; Tapiovaara M, 1997).

2.3. Experimental measurement of absorbed dose

Experimental measurements were carried out to verify the accuracy of the MC simulations. For these measurements of the absorbed dose, a 0.23 cm³ FC23C Wellhöfer Farmer type ion chamber was placed inside a CIRS IMRT homogeneous phantom. k_{ch} was assumed to be 1 for this chamber. The ion chamber was displaced 3.0 cm anterior from the midline to mimic the prostate location. The CIRS phantom had an elliptical cross section 20 cm thick and 30 cm wide.

The phantom was irradiated with a single projection kV beam from the AP direction (0° imager angle) and source-axis distance of 100 cm for field sizes of 4 cm × 4 cm to 8 cm × 8 cm in 0.5 cm × 0.5 cm increments, giving a total of nine measurements. The characteristics of the experimental kV beam (delivered with a Varian OBI) were 125 kV, 200 mA and 50 ms. For experimental verification, a higher mAs setting (i.e. 10 mAs) was used compared to that used in the simulations (i.e. 1.04 mAs). The higher mAs setting was used in the experimental verification to achieve a high signal-to-noise ratio. The absorbed dose is directly proportional to the current-time product. Hence, the experimentally measured absorbed dose could be scaled down to provide an equivalent comparison to the MC calculated absorbed dose (at 1.04 mAs). In the MC simulation results presented in the remainder of this paper, the value of 1.04 mAs was used consistently.

The experimental absorbed dose was determined using the AAPM TG-61/IPEMB protocol (equivalent at these energies) (Aukett *et al.*, 2005; Ma *et al.*, 2001; Klevenhagen *et al.*, 1996). To determine the whole organ absorbed dose from the point of measurement, we assumed that the absorbed dose to the whole prostate was the same as the measured point absorbed dose. This assumption was justified because the prostate is relatively small and the treatment is typically from multiple angles.

For the calculated comparisons, the PCXMC MC phantom width and thickness were set to match that of the CIRS phantom. The x-ray beam characteristics were also set to be the same as that of the experimental parameters (i.e. 10 mAs instead of 1.04 mAs). These changes were performed so that a fair comparison could be made between the experimental and calculated values.

Results

First, the effective dose per projection as a function of imager angle and field size was determined. The average angular effective dose was then calculated. The relationship between the average angular effective dose with treatment duration and imaging frequency was then shown. Following this, the experimental verification of the MC calculations was provided.

3.1. Effective dose as a function of imager angle

The parameters discussed above were input into PCXMC. For the standard phantom, the effective dose per projection was computed as a function of kV source angle (0° – 360°) for a specific field size. Figure 1 shows the curve computed for the standard phantom at a field size of $6\text{ cm} \times 6\text{ cm}$ and also for the AP, PA and LAT values computed by Le Heron *et al.* (Heron, 1992). The variation in effective dose with angle reaches approximately a factor of 6.3. An average angular dose was then calculated for the standard phantom. This is the effective dose averaged over all angles. Also from figure 1, the AP (0°) projection has a higher effective dose than the PA (180° or -180°) projection. This is because the AP projection delivers more effective dose to the testes (a very radiosensitive organ) and bladder. The PA projection delivers a dose to these organs that is partially shielded by the posterior muscles and lower spine. The left projection (90°) of the phantom receives slightly more effective dose than the right projection (-90°) because of the presence of the spleen and pancreas on the left side. Overlaid on figure 1 are the relevant data points from the Le Heron *et al.* (Heron, 1992) study which have been reproduced in TG75 (Murphy *et al.*, 2007).

3.2. Effective dose as a function of field size

Figure 2 displays a linear increase in effective dose per projection with field size, for field sizes varying from $4\text{ cm} \times 4\text{ cm}$ to $8\text{ cm} \times 8\text{ cm}$.

3.3. Effective dose as a function of treatment duration

To compute the effective dose as a function of treatment duration, a dose representing the general case incorporating all of the non-treatment parameters was determined. The effective

dose was averaged over all angles to give the angular average effective dose, giving a value of 0.9 μSv per projection.

For KIM, a pretreatment arc is required for each fraction. Because KIM converts 2D projection positions into 3D positions, a 3D probability density function (pdf) needs to be built from the pretreatment arc (Poulsen *et al.*, 2008). Poulsen *et al.* demonstrated that a 20 s pretreatment arc is sufficient to build a prostate motion pdf that achieves sub-mm accuracy (Poulsen *et al.*, 2009). For each fraction, a 20 s pretreatment imaging arc was added for all of the KIM treatment durations.

To compute the effective dose per fraction in table 4, examples of treatment durations are needed. These data are obtained from a study by Oliver *et al.* (Oliver *et al.*, 2009). For conventional fractionation, the treatment durations are typically 187.2 minutes for IMRT (average of 5 and 9-field) and 85.8 minutes for IMAT (average of single and double arc). For SBRT, the treatment durations are 78.0 minutes for IMRT and 36.0 minutes for IMAT.

3.4. Experimental measurement of absorbed dose

Table 3 compares the experimental and simulation results on absorbed dose to the prostate for various field sizes. The percentage differences for all measurement points are less than 7% and the two sets of data agree to within the experimental uncertainty (AAPM TG-61 (Ma *et al.*, 2001) estimates a one-sigma uncertainty of 3.6–4.7%).

Discussion

In this study we calculated the effective dose as a function of imaging angle and field size. To verify the accuracy of the MC calculations, we also measured the absorbed dose experimentally. Here our results are discussed in the context of other work.

For effective dose against imaging angle in Figure 1, it is evident that the calculations of Le Heron *et al.* agree reasonably well with that of PCXMC. However, Le Heron's AP dose is slightly lower than that of PCXMC. We believe this is due to the difference in prostate tissue weighting factors utilized by Le Heron *et al.* and PCXMC. Le Heron *et al.* conducted their study in 1992 and hence used the tissue weighting factors obtained from ICRP Publication 60 (1991) whereas PCXMC uses the tissue weighting factors from ICRP Publication 103 (2007). The prostate tissue (considered 'remainder') weighting factors are 0.05 and 0.12 respectively. The prostate, which is displaced anteriorly, will receive more AP dose. Combined with the fact that PCXMC uses a higher tissue weighting factor for the prostate, we believe this explains the discrepancy in effective dose seen here.

For effective dose against field size in Figure 2, we note that as the field size increases, two factors can potentially influence the relationship between the effective dose and field size. First, the organs surrounding the beam central axis have different tissue weighting factors and are situated in a complex arrangement. Second, the non-uniformity of the x-ray beam across its area will be more significant for larger field sizes. Our results show, however, that these factors have a negligible effect on the linear relationship between effective dose and field size in the range of the field sizes considered.

For the comparison between the measured and calculated absorbed dose in table 3, we note that the measured values are systematically higher than the calculated values by 6%. Although this is considered reasonable agreement in the context of this work, the difference can be explained as follows. Firstly, the CIRS phantom had its water equivalency only validated at MV energies. This does not ensure water equivalency at kV energies. Secondly, the small air gap between the CIRS phantom insert and the Farmer ion chamber used to measure the dose cannot be accurately simulated with PCXMC. The presence of the air gap results in increased absorbed dose to the ion chamber. Both these factors contribute to the systematic difference.

We also calculated the effective dose from the KIM real-time localization modality for various treatment scenarios. The average effective dose was 0.9 μ Sv per projection for a field size of 6 cm $\times$ 6 cm. This value was used to calculate the effective dose as a function of treatment type, fractionation and treatment duration at an imaging frequency of 1 Hz, yielding effective doses ranging from 2–12 mSv. In table 4 the effective dose from the KIM modality is compared to another real-time localization modality, i.e. the Navotek radioactive implant, and an IGRT method, i.e. CBCT image guidance. The Navotek real-time tumour position monitoring system involves the implantation of a 100 μ Ci Ir-192 radioactive marker into the patient's prostate (Shchory *et al.*, 2010). Once implanted, this marker is not removable for the patient's lifetime. Neustadter *et al.* estimated, using MC simulations, that the effective dose delivered by this radioactive implant throughout the patient's lifetime is 64 mSv (assuming no retreatment) (Neustadter *et al.*, 2010). In table 4, the KIM doses for all treatment scenarios considered were lower compared to the Navotek Ir-192 radioactive implant dose.

For reference purposes, a Varian OBI pelvic CBCT scan dose is included in table 4. This was measured by Hyer *et al.* using a fiber-optic coupled dosimetric system (Hyer *et al.*, 2010). The scan parameters were as follows: 125 kV, 800 ms, 13 mA and 655 projections. This CBCT scan delivered an effective dose of 4.3 mSv. For conventionally fractionated and SBRT IMRT and IMAT, the KIM dose is lower compared to a pelvic CBCT scan dose for an entire treatment. This is because the number of projections, field size and imaging frequency are lower than that of CBCT scans. Also, note that a CBCT scan is not a real-time localization tool. Its dose values were listed in order to provide a meaningful comparison with KIM dose.

As noted by He *et al.*, (He *et al.*, 2010) there is a variation in dose due to projection angle - more than a factor of 6 in this study - and further dose reductions could be obtained by using x-ray tube current modulation.

It is expected that as clinical dose rates increase, the dose from KIM will decrease. With the advent of 2400MU/min linacs (Scorsetti *et al.*, 2011), it is expected that treatment delivery times will be reduced. Furthermore, as treatment times are reduced, the dose from kV guidance will also be reduced. It is acknowledged that the increased dose rates will increase the MV scatter and hence reduce the image quality of the kV images. This would require the kV exposure parameters and hence dose to be increased for sufficient image quality. However, techniques to circumvent this include specifying control points which pause the

MV beam intermittently allowing for kV acquisition to occur during no MV irradiation (Ling *et al.*, 2011).

The clinical feasibility of KIM for prostate cancer has recently been demonstrated, resulting in a position monitoring accuracy of 0.46 mm on average (Ng *et al.*, 2012). The quantification of the dose from this modality, along with strategies to reduce this dose, are important prior to the broader clinical use of the KIM modality.

Conclusion

In this study, we quantified the additional dose delivered for prostate radiotherapy with Kilovoltage Intratreatment Monitoring (KIM). For all treatment scenarios considered here (i.e. conventionally fractionated and SBRT IMRT and IMAT), the KIM modality delivered less imaging dose than the Navotek radioactive fiducial implant and a standard SBRT CBCT pretreatment scan over an entire treatment regime.

The effective dose varied significantly by more than a factor of 6 with imager angle. For a field size of 6 cm × 6 cm, the effective dose was found to be 0.9 µSv, averaged over all projection angles. The KIM effective dose increases linearly with field size, treatment duration and imaging frequency.

In conclusion, this novel KIM modality has been shown to deliver lower effective imaging dose than other clinically accepted prostate intratreatment guidance modalities.

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Appendix

Table 5

Summary of imaging dose studies for gantry-mounted x-ray IGRT procedures.

System	Method	Body Region	Dimensions	Internal Filter	External Filter	kVp	mAs	Results
Islam <i>et al.</i> (Islam <i>et al.</i> , 2006) ^a	Ion chamber/MOSFET measurements	32 cm diameter body phantom 16 cm diameter head phantom	Detector sensitive area = $41 \times 41 \text{ cm}^2$ SDD = 155cm	2mm Al 0.1mm Cu	None	120	660	Body: 16 mGy (center) 23 mGy (surface) No calculations of effective dose. Head: 30 mGy (center) 29 mGy (surface). No calculations of effective dose.
Amer <i>et al.</i> (Amer <i>et al.</i> , 2005) ^a	Measured	Pelvic	Not stated	2mm Al 0.1mm Cu	Not stated	130	$480 = (1.2 \text{ mAs} \times 400 \text{ projections})$	25 mGy. No calculations of effective dose.
Cossmann <i>et al.</i> (Cossmann <i>et al.</i> , 2005) ^a	Measured	16 cm cylindrical head phantom	48 cm FOV 17 cm scan length	2.8 mm Al	Bowtie filter	125	Low dose: $246.4 = (0.4 \text{ mAs} \times 616)$ Standard dose: $1232 = (2.0 \text{ mAs} \times 616)$	15 mGy (center and surface). No calculations of effective dose. 74 mGy (center) 72 mGy (surface). No calculations of effective dose
Endo <i>et al.</i> (Endo <i>et al.</i> , 1999) ^a	Prototype TLDs	Rando phantom	Not stated	Not stated	Not stated	120	720	11 mSv (head) 25 mSv (chest)
Hyer <i>et al.</i> (Hyer <i>et al.</i> , 2010)	Elekta XVI and Varian OBI Fibre-optic coupled dosimetry system	Phantom Pelvis	Not stated	2.8 mm Al	Half bowtie	125	$680 = (1.04 \times 655)$	4.34 mSv

	System	Method	Body Region	Dimensions	Internal Filter	External Filter	kVp	mAs	Results
Gu <i>et al.</i> (Gu <i>et al.</i> , 2008)	Elekta XVI and Varian OBI	MCNPX	VIP-Man Phantom	SDD = 150 cm	4 mm Al	None	125	1350 = (2 mAs × 675)	6.25 mSv (per prostate scan)
He <i>et al.</i> (He <i>et al.</i> , 2010)	Generic Cone Beam Imager and C Arm	PCXMC	Mathematical Phantom Chest and abdomen only	SDD = 93 cm	Various	None	80, 120, 140	Various	Effective dose as function of projection angle
Spezi <i>et al.</i> (Spezi <i>et al.</i> , 2012)	Elekta XVI	EGSnrc/BEAMnrc	Pelvis Chest Head and Neck	Not stated	2 mm Al 0.1 mm Cu	Various	120 120 120 100	1.6 per projection 1.6 per projection 0.1 per projection	20–60 mGy. No calculations of effective dose. 12–34 mGy. No calculations of effective dose. 1–2 mGy. No calculations of effective dose.
Ding <i>et al.</i> (Ding <i>et al.</i> , 2008)	Varian OBI	BEAMnrc/DOSXYZnrc + Experimental calibration	Pelvis Chest Head and Neck	Protocol default	2.8 mm Al	Protocol default	125	2.0 per projection	Organ specific. 40 mGy to prostate in abdominal scan. No calculations of effective dose.
Ding <i>et al.</i> (Ding and Coffey, 2009)	Varian OBI	DOSXYZnrc + Experimental calibration	Head and Neck Abdominal Chest	Protocol default	2.8 mm Al	Protocol default	125	2.0 per projection	Organ specific. 30 mGy to prostate in abdominal scan. No calculations of effective dose.
Chow <i>et al.</i> (Chow <i>et al.</i> , 2008)	Elekta XVI	BEAMnrc/EGSnrc (uncalibrated to absolute dose)	Pelvis	Protocol default	2mm Al 0.1mm Cu	Protocol default	100, 120, 130, 140	Not relevant as no calibration used.	Various regions of interest. No calculations of effective dose.
Crocker <i>et al.</i>	Varian OBI	Ion chamber	CIRS Phantom	Protocol default	2.8 mm Al	None	75 120	10 per projection 1.04 per projection	Absorbed dose for

System	Method	Body Region	Dimensions	Internal Filter	External Filter	kVp	mAs	Results
(Crocker <i>et al.</i> , 2012)								various treatment scenarios. No calculations of effective dose. Absorbed dose for various treatment scenarios. No calculations of effective dose.
Le Heron <i>et al.</i> (Heron, 1992)	Monte Carlo	Mathematical phantom		Various	None	Various	Not specified.	Lookup table of conversion coefficients (mSv/Gy cm ²) for various voltages, filtrations and body regions.
Our study	Varian OBI PCXMC Monte Carlo	Reduced field prostate	SDD = 180cm	2.8 mm Al	None	125	1.04 per projection	Effective dose as a function of angle, patient size and treatment type, treatment duration and imaging frequency

^a All the above measurements are consistent with each other when adjusted for technique, geometry and scan protocol (Murphy *et al.*, 2007).

Figure 1.

MC calculated angular variation of single projection effective dose for the standard adult male for a field size of 6 cm × 6 cm. The average angular dose (shown in brackets in the legend) is the effective dose averaged over all angles. The Le Heron *et al.* (Heron, 1992) effective dose points (black triangles) are also plotted.

Figure 2.
The angular average effective dose per projection as a function of field size.

Table 1

Parameters used in this study divided into two categories: x-ray beam and treatment parameters.

Parameters	
X-ray beam parameters	
Filtration	2.8 mm Al (no bowtie filter)
Anode Angle (deg)	14
Tube Voltage (kV)	125
Tube Current (mA)	80
Exposure Time (ms)	13
Current-Time Product (mAs)	1.04
kV Source Angle in IEC (deg)	−180, −170, ..., 0, ..., 170, 180
Source Isocenter Distance (cm)	100
Source Detector Distance (cm)	180
Field Size at Isocenter (cm × cm)	4×4, 4.5×4.5, ..., 8×8
Treatment parameters	
Treatment type	IMRT and IMAT
Fractionation	Conventional fractionation and SBRT
Treatment duration (s)	Depends on treatment type and fractionation
Imaging frequency (Hz)	0.1, 0.5 and 1

Table 2

Parameters of the modified Cristy and Eckerman²⁵ adult standard phantom in PCXMC.

Parameter	Value
Age	Adult
Mass (kg)	73.2
Total height (cm)	178.6
Trunk thickness (cm)	20.0
Trunk width (cm)	34.4

Table 3

Comparison of experimentally measured and MC calculated absorbed doses to the prostate for various field sizes. The value after the $\pm$ symbol indicates the standard error of the mean.

Field Size (cm $\times$ cm)	Measured absorbed dose (mGy)	Calculated absorbed dose at 10 mAs (mGy)	% Difference (between measured and calculated)
4 $\times$ 4	0.44 $\pm$ 0.03	0.42 $\pm$ 0.01 ^a	6.1
5 $\times$ 5	0.50 $\pm$ 0.04	0.47 $\pm$ 0.01	5.8
6 $\times$ 6	0.56 $\pm$ 0.04	0.52 $\pm$ 0.01	6.3
7 $\times$ 7	0.59 $\pm$ 0.04	0.55 $\pm$ 0.01	6.3
8 $\times$ 8	0.66 $\pm$ 0.05	0.61 $\pm$ 0.02	6.4

^aThis statistical uncertainty can be reduced by increasing the number of MC simulation photons.

Table 4

Effective dose due to kV intratreatment monitoring (for a field size of $6\text{ cm} \times 6\text{ cm}$ and imaging frequency of 1 Hz) in several treatment scenarios compared against the Navotek radioactive implant and daily CBCT image guidance. The duration per fraction times are adapted from Oliver *et al.* (Oliver *et al.*, 2009) with an additional 20 s added to each fraction for pretreatment imaging.

Intratreatment guidance modality	Treatment type	Fractionation	Duration per fraction (s)	Number of fractions for total treatment	Effective dose for entire treatment (mSv)
kV intratreatment monitoring	IMRT	Conventional	307	39	10.2
		SBRT	951	5	4.1
	IMAT	Conventional	151	39	5.0
		SBRT	444	5	1.9
Navotek Ir 192 implant Pelvic CBCT pretreatment scan ^a	Independent	Independent	Independent	Independent	64
	Independent	Conventional	Independent	39	169.7
		SBRT	Independent	5	21.7

^a Although CBCT is not a real-time localization modality, it is included for reference as it is a commonly used image-guided radiation therapy (IGRT) method for prostate cancer.