Comparative study of a Low Z Cone Beam Computed Tomography system

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Abstract

Computed tomography images have been acquired using an experimental (low Z insert) megavoltage cone beam imaging system. These images have been compared with standard megavoltage and kilovoltage imaging systems. The experimental system requires a simple modification to the 4MeV electron beam from an Elekta Precise linac. Low energy photons are produced in the standard medium Z electron window and a low Z carbon electron absorber is used to remove any remaining electrons from the source. A detector sensitive to diagnostic x-ray energies is also employed. Quantitative assessment of CBCT contrast shows that the Low Z imaging system is orders of magnitude superior to a standard 6 MV imaging system. CBCT data with the same contrast to noise ratio as a kilovoltage imaging system (0.147 cGy) can be obtained in doses of 14.72 cGy and 239.53 cGy for the experimental and standard 6MV systems respectively. Whilst these doses are high for everyday imaging, qualitative images indicate that kilo-voltage like images suitable for patient positioning, can be acquired in radiation doses of 1-8 cGy with the experimental Low Z system.

1. Introduction

Radiotherapy requires high doses of radiation to be delivered accurately in both position and intensity. With the now almost ubiquitous use of new image guided and delivery techniques such as in-room imaging and intensity modulated radiotherapy, radiation can be delivered to millimetre accuracy and to a host of complex tumour geometries. Further to this, the recent mainstream use of image guided radiotherapy, most of which use gantry mounted kilovoltage systems allow for accurate daily patient setup (Amer et al. 2007; Jaffray et al. 2002). These cone beam computed tomography systems (CBCT) allow for a direct comparison between a 3D volume taken at treatment and at the radiotherapy planning stage. These systems however require complex add-on systems and do not allow for imaging through the beams eye view. Whilst several other non-ionising techniques for ensuring accurate patient positioning before and during treatment exist, the 'x-ray' remains by far the most widely used modality (Evans 2008).

Several authors have sought to address this issue, even before the concept of gantry mounted kilovoltage systems by implementing megavoltage cone beam imaging systems (MV-CBCT). Early implementation of MV-CBCT (Simpson et al. 1982) was concerned with using the 3D volume for patient planning due to the lack of dedicated CT scanners for treatment planning. It was also realised that megavoltage CBCT could be used for patient positioning but in both cases the systems required large radiation doses to produce acceptable images. Further research into MV-CBCT, primarily through improvement in detector technology (Lewis et al. 1992; Mosleh-Shirazi et al. 1998) reduced the radiation dose to acceptable levels (circa 10 cGy) enabling regular patient imaging (Pouliot et al. 2005).

In this paper we report on an experimental imaging system utilising a megavoltage beam from a standard radiotherapy linear accelerator. We have previously shown (Roberts et al. 2008) improved planar imaging using a system employing a thin medium atomic number (Z) electron window, a low Z electron absorber and an imaging panel sensitive to diagnostic energies. Here Cone beam computed tomography with this system is reported. Several

authors have implemented similar systems utilising Low Z targets (Faddegon et al. 2008; Flampouri et al. 2002; Flynn et al. 2009; Pouliot et al. 2005; Robar et al. 2009) and have shown vast improvements over treatment beam imaging. Both planar and CBCT acquisitions have been reported but not directly compared with the systems that they would replace. In all cases there has been no definitive comparison with kilo-voltage systems. Whilst megavoltage imaging systems utilising electron beam energies of circa 4-5 MeV would not be expected to achieve kilovoltage quality they may offer a simpler and cheaper alternative with an acceptable dose.

2. Method

The experimental system is shown in Figure 1 and utilises a standard Elekta Precise linac (Elekta Ltd, Crawley, UK) which was modified to include a carbon electron absorber in the high energy collimator port. By running the machine at the lowest electron energy, x-rays are produced in the thin medium Z vacuum window and electrons emerging from the window are subsequently absorbed by the carbon absorber. 70% of the photons at isocentre are produced in the nickel window, whilst the other 30% are produced in the absorber (Roberts et al. 2008). For this work the accelerator was operated in the 4 MeV high dose rate electron (HDRE) mode. Previous Monte Carlo simulations have shown that this beam can be modelled as a truncated Gaussian with a mean energy of 5.6 MeV, a full width at half maximum energy of 1.9 MeV and lower and upper limits of 3.8 and 6.2 MeV respectively. (Roberts et al. 2008).

Figure 1- Experimental imaging system implemented on an Elekta Precise linac utilising the standard Elekta medium Z electron window and low Z electron absorber

In this work CBCT imaging is compared for 4 different systems listed in Table 1. For the kV/VXI system the tube energy and choice of bowtie filter is taken from standard practice depending on the body site to be imaged.

System Name (Beam/ panel)	Description
6MV/iViewGT	Standard Elekta megavoltage imaging system utilising the 6MV treatment beam and an amorphous silicon imaging panel with a gadox scintillator screen
6MV/XVI	Standard Elekta megavoltage imaging system utilising the 6MV treatment beam and an amorphous silicon imaging panel with a CsI columnar scintillator screen
100kV/XVI 120kV/XVI 120kVBow/XVI	Standard gantry mounted kilovoltage imaging system (XVI, Elekta Ltd) employing a kilovoltage x-ray tube (operating at 100 or 120 kVp) and an amorphous silicon imaging panel with a CsI columnar scintillator screen. The 120kVBow/XVI system includes a bow tie filter for scanning the pelvis region.
LowZ/XVI	The experimental system being tested with a low Z insert and an amorphous silicon imaging panel with a CsI columnar scintillator screen.

Table 1- Imaging systems under investigation. Gadox is gadolinium oxysulphide. CsI is caesium iodide and XVI is X-ray Volume Imaging

2.1 Cone beam acquisition method

CBCT acquisitions for the linac generated beams (LowZ and 6MV) were obtained using a rotating table with the linac gantry at 90°. For the kilovoltage system, XVI 4.5 (Elekta Ltd, Crawley) was used to acquire the projection data using a standard acquisition sequence i.e. the gantry rotated whilst images were obtained. In all cases the imaging panels ran in a gated frame read mode. For the kilovoltage system (XVI) this is the standard acquisition method whereby the x-ray tube is pulsed and the panel is subsequently read out. For the megavoltage systems (6MV, LowZ) a gating interface was used to turn off the radiation pulses during panel read out (figure 2). This was achieved by holding off the current pulse to the magnetron (and by effect the pulse to the electron gun) during panel readout.

Figure 2 - Gated frame read mode for megavoltage beams utilising the XVI panel. Figure shows the detector control board (DBC) signal to the magnetron enable circuitry of the linac and the magnetron current signal (Mag I) which is present when the linac is gated on. Note the apparent variation in the Mag I pulse amplitude due to aliasing effects of the digital oscilloscope.

For the megavoltage beams the radiation on time per frame was set to 200 ms. Adjustment of the dose per frame was achieved by decreasing the pulse repetition frequency (PRF) or reducing the electron gun current. Tissue phantom ratio measurements were performed to ensure that the beam energy was not affected by the adjustment of the electron gun current. The read out times were 434.616 ms and 142.560 ms for the iViewGT and XVI panels respectively. Note that for the megavoltage beams the radiation off time was longer than the frame read time by 2.24 ms. This arose from delays in the gated frame read system.

All projection data was reconstructed using an in-house CBCT reconstruction program (Cone.exe, Institute of Cancer Research) which is based on the Feldkamp algorithm. Prior to reconstruction all projections except for the catphan spatial resolution phantom were down-sized from 1024x1024 pixels to 512x512 pixels. The reconstruction parameters are detailed in Table 2. Note that no scatter correction algorithms were used in this work. The catphan spatial resolution phantom had higher resolution projection and reconstruction images so that the maximum spatial resolution of the systems could be determined. All other phantoms were reconstructed with settings that would correspond to normal clinical practice. The benefits of not reconstructing at the highest possible settings is that reconstruction times and storage space are significantly reduced whilst not affecting patient positioning accuracy.

Parameter	Description	Setting
Reconstruction Size	Number of pixels used in each of the x, y and z directions	256 (head, pelvis, catphan contrast) 512 (catphan spatial resolution)
Reconstruction dimensions	Determines the reconstruction limits Axial Radius	-12.5 to 12.5 cm (all phantoms) 12.5 cm (head and catphan) 17 cm (pelvis)
Projection filter	Determines which image filter will be used on the projections prior to reconstruction. The number donates the size of the filter	5 element median window filter
Reconstruction filter	Determines which filter is used in the reconstruction	Hamming

Table 2- Reconstruction parameters

Due to the limited field of view of the imager two imaging geometries were used to gather projection data for small (<20 cm) and large phantoms (>20 cm). These are shown in Figure 3. The first geometry only requires projection data to be acquired over 200° and is labelled ‘short scan’ whilst the second geometry required a full 360 degrees and is labelled ‘full scan’.

Figure 3- Imaging geometries (a) short scan and (b) full scan

The field size used for the linac based 6 MV beams was 20x20 cm² for phantoms up to 20 cm in diameter (catphan and head phantoms below) and 20x20 cm², offset by 7 cm for phantoms greater than 20 cm in diameter. The kV/XVI systems used the S20 and M20 collimators for small (<26 cm diameter) and large (> 26cm diameter) phantoms respectively.

2.2 Dosimetry

The weighted computed tomography dose index (CTDI_w) is a popular value quoted for kilovoltage cone beam CT systems and standard diagnostic CT systems (both spiral and axial). It can be described by equation (1) where D_N, D_E, D_W, D_S and D_C are the doses at the north, east, west, south and central points respectively. A schematic of a CTDi phantom is shown in Figure 4 indicating the dose measurement regions. Two CTDi phantoms were used with diameters of 16 cm and 32 cm. The use of two different phantoms allows the quantification of dose to different regions of the body e.g.16 cm corresponds to the head region and the 32 cm phantom to the pelvis.

$$CTDI_w = \left(\frac{2}{3} * \sum_{All_angles} \left(\frac{D_N + D_E + D_W + D_S}{4} \right) \right) + \left(\frac{1}{3} * \sum_{All_angles} D_C \right) \quad (1)$$

Figure 4- Schematic diagram of CTDI phantom showing the central and peripheral (North, East, South and West) regions

In equation (1) the peripheral doses are weighted to a greater extent than the central dose. This is because dose falls off quickly from the surface to the centre of the phantom with a kilovoltage beam. Also note from Figure 4 that a 10 cm long ionisation chamber is normally used to obtain the dose along the length of the scan. However the use of a 10 cm long ionisation chamber and a CTDI phantom that is shorter than the CBCT field size results in underestimates of the CTDI. It has been reported that central dose values are underestimated by as much as 30% and peripheral doses by 8% due to the lack of scatter from the side of the phantom (Amer et al. 2007). This method has therefore been modified in this work by the use of a Farmer chamber in the CTDI phantom to measure the central slice CTDI. Because the phantoms were not designed to accommodate a Farmer ionisation chamber, water was introduced into the measurement hole with the chamber to remove any air gaps. The use of a Farmer chamber is also appropriate here as the ones available for this study were calibrated for use in kilo-voltage and megavoltage beam qualities.

As well as reporting the weighted central slice CTDI for both phantoms ($CTDI_{cw16}$ and $CTDI_{cw32}$), the dose indices C16 and C32 are used to indicate the dose at the central points of the two CTDI phantoms. These values are quoted as well as the $CTDI_{cw}$ as megavoltage CT images may have higher central than peripheral doses which could make the comparison of $CTDI_{cw}$ measures for kV and MV beams misleading.

If the object being scanned was less than 20 cm in diameter and used a 'short scan' detector geometry then the C16 and $CTDI_{cw16}$ indices were measured. However if the object is greater than 20 cm the C32 and $CTDI_{cw32}$ indices were used.

For all megavoltage beams the CTDI was calculated by using a BEAMnrc (Rogers et al. 1995; Walters B et al. 2004) Monte Carlo model of the x-ray beams (Roberts et al. 2008) and impinging these on a model of a CTDI phantom, which was simulated in DOSXYZnrc (Walters B et al. 2004). The dose distribution in the phantom was then obtained and summed over all gantry angles to determine the relative doses at the 5 measurement points. The relative doses were then normalised based on a measurement using a Farmer chamber (PTW, Germany) and Unidos electrometer (PTW, Germany) at the central location. All megavoltage dosimetry was conducted in accordance with TRS-398 (Andreo P et al. 2004).

For the kilovoltage beam the dose at the surface of the phantom was determined in accordance with the AAPM TG-61 protocol for kilovoltage x-ray beam dosimetry (Ma et al, 2001). Air kerma was measured in air using an NE2571 farmer chamber and NE electrometer. Backscatter and mass attenuation coefficient parameters were determined at the relevant source to surface distance (SSD), field size and half value layers (HVLs) of the two XVI beams (100 and 120 kVp). Subsequently the dose at the centre of the phantom was obtained by simulating the depth dose curve of the kilovoltage beams in BEAMnrc with input x-ray spectra obtained from SpekCalc (Poludniowski et al. 2009). The dose distribution for the kilovoltage beams in the CTDI phantoms were determined using Monte Carlo as described above for the megavoltage beams.

2.3 Phantom images

For quantitative comparison of the different imaging systems the Catphan phantom (The phantom laboratory, Salem, USA) was used. This phantom is cylindrical in shape and has two main inserts. One contains a series of line pairs ranging from 1 to 21 lines pair per mm. These were used to assess the spatial resolution of the systems by visually determining which line pairs could be resolved. The other sections contain inserts of different materials and densities. In this study the contrast to noise ratio was determined between the polytetrafluoroethylene (PTFE) insert (density 2.16 g.cm⁻³) and the water equivalent background (density 1.02 g.cm⁻³) using equation (2) where $\bar{I}_{background}$ is the average pixel value of the region surrounding the object, $\bar{I}_{object}$ is the average pixel value of the object and $\sigma_{background}$ is the standard deviation of the background region.

$$CNR = \frac{\bar{I}_{background} - \bar{I}_{object}}{\sigma_{background}} \quad (2)$$

For qualitative comparison, head and pelvis anthropomorphic phantoms (RSD, Long Beach, USA) were imaged.

3. Results

3.1 Contrast to noise ratio

Figure 5 shows the relationship between CNR^2 and the central dose for the LowZ/XVI system and standard imaging systems determined from the catphan phantom. CNR^2 is plotted against dose as this should be directly proportional to the imaging dose (K.Faulkner and B M Moores 1984).

Figure 5-Dose vs. contrast to noise ratio for experimental and standard radiotherapy imaging systems. Symbols denote experimental measurements whilst lines show the straight line fit through these points with the y-axis intercept set to 0. C16 is the dose deposited at the centre of a 16 cm wide CTDI phantom

Through extrapolation of the straight line fits in Figure 5 the dose required to get the same CNR^2 value as the 100kV/XVI system was ascertained. The imaging setting for the 100kV/XVI system was the standard head preset on the XVI system (10 ms, 10 mA per frame and a total of 360 frames over 200 degrees). The results of this analysis are presented in Table 3.

Beam	CNR^2	Dose (at centre of 16cm phantom) /cGy	Ratio of dose to 100kV/XVI
100kV/XVI	182.14	0.147	1.00
120kV/XVI	182.14	0.201	1.37
LowZ/XVI	182.14	14.713	100.10
6MV/XVI	182.14	57.880	393.74
6MV/iViewGT	182.14	239.527	1629.44

Table 3 - Dose required to obtain the same CNR value as the kV/XVI systems. The contrast to noise ratio is measured between the Catphan phantom background (~water) and the PTFE insert.

A wide range of doses are required to get the same image quality as the kVI/XVI system, none of which are acceptable for everyday use. The 120kV/XVI system requires more dose than the 100kV/XVI system due to the size of the catphan phantom. If a larger phantom

where used then the 120kVp/XVI system would give the lowest dose for a given image quality. This is the reason why the 120kVp is used clinically for imaging treatment sites such as the pelvis and torso. The 6MV/iViewGT dose is greater than a single therapy fraction and highlights how important the use of kilovoltage x-rays is for imaging. Using the XVI panel with the 6MV beam requires a factor 4 less dose but is still unacceptable. This dose saving results from the more quantum efficient nature of the CsI based panel and its sensitivity to lower energy photons that lead to higher contrast images.

The LowZ/XVI beam on the other hand requires orders of magnitude less dose than the 6MV system resulting in a dose approaching that of a few planar portal images. Whilst 14 cGy is likely to be high for regular imaging, it must be noted that we do not necessarily need the spatial resolution and contrast of the kV/XVI systems to perform patient positioning. Images corresponding to such doses (<10cGy) are shown in Figure 7 and Figure 8 and clearly show that acceptable images can be obtained at an acceptable radiation dose.

3.2 Spatial resolution

Spatial resolution was assessed by using the line pair section of the catphan phantom. The line pair slice for four imaging systems can be seen in Figure 6. For this test large imaging doses and high resolution reconstructions were used for all systems to determine the inherent resolution of the systems and not necessarily the resolution achievable under normal imaging doses. The reconstruction algorithm and collimator size also affect the maximum achievable resolution.

Figure 6 - Catphan spatial resolution image reconstruction for (a) 6MV/iViewGT, C16=30.91 cGy (b) LowZ/XVI, C16=26 cGy (c) 100kV/XVI (40 ma, 40 ms), C16=2.63 cGy and (d) 120kV/XVI (40 ma, 40 ms), C16=4.62 cGy. Window levels are set to levels where the maximum number of line pairs are visible for each system.

Superior spatial resolution can be seen for the kV-XVI/XVI system with the first 10 line pairs resolvable, whilst the first 7 are visible for the LowZ/XVI system. Only 2 are just visible for the 6MV/iViewGT system owing to the very noisy images.

3.3 Phantom Images

Figure 7 and Figure 8 show the CBCT reconstructions of humanoid anthropomorphic phantoms for the LowZ/XVI, 6MV/iViewGT and kV/XVI systems. The doses for the 6MV/iViewGT and LowZ/XVI systems were chosen to be below 10 cGy (a figure considered practicable). The kV-XVI/XVI dose values were the results of the standard head and pelvis protocols on the XVI system.

Previously it had been shown that large doses would be required to match the kV-XVI/XVI image quality. However, Figure 7 and Figure 8 show that acceptable image quality can be obtained for doses less than 10 cGy. Figure 7 shows a series of head phantom reconstructions for the various beams and demonstrates that with a dose of around 1 cGy images of a head can be obtained using the LowZ/XVI system that could be suitable for patient positioning (7b). The 6MV/iViewGT system (7a) is shown to be worse than all the other systems whilst the kilovoltage system (100kV/XVI) (7d) shows superior image quality for a substantially smaller radiation dose as expected.

Images of the pelvis phantom in Figure 8 again show acceptable images for the LowZ/XVI system with a dose of around 7 cGy (8b). This dose is equivalent to 2 or 3 portal images and therefore would be acceptable as a method for patient positioning. Images with the 6MV/iViewGT system (8a) require doses in excess of 10 cGy whilst producing reconstructions considerably worse than the low Z system. Once again the kilovoltage system (8c) shows superior image quality at a reduced dose, highlighting the importance of low energy kilovoltage photons in the production of low dose, high contrast images.

It must also be noted here that two dose indices have been reported with the proceeding images. The dose to the centre of a phantom (i.e. the C16 and C32 indices) is higher than the $CTDI_{cw}$ for the LowZ and 6MV beams whilst the opposite is true for the kilovoltage system. This is due to the sharp drop off of dose with depth for kilovoltage systems. It is not clear which dose index would be most useful and this would be a clinical decision based on whether the maximum dose or the average dose to the whole body were considered the most important factor. Due to the high doses that may be involved with megavoltage CBCT systems, recent studies have investigated taking the imaging dose into account when planning a patient's treatment (Flynn et al. 2009). This would be increasingly important if imaging doses of greater than 10 cGy were given to improve soft tissue visualisation.

Figure 7 - 3D reconstructions of a Rando-Alderson head phantom using a short scan geometry for (a) 6MV/iViewGT ($C16=6.30$ cGy, $CTDI_{cw16} = 6.59$ cGy), (b) LowZ/XVI ($C16 = 1.21$ cGy, $CTDI_{cw16} = 0.91$ cGy), (c) LowZ/XVI ($C16 = 6.63$ cGy, $CTDI_{cw16} = 5.01$ cGy), (d) 100kV/XVI normal scan ($C16=0.147$ cGy, $CTDI_{cw16} = 0.166$ cGy). The horizontal lines in the coronal and sagittal slices are due to the segmented nature of the phantom.

Figure 8 - 3D reconstructions of a Rando-Alderson pelvis phantom for (a) 6MV/iViewGT ($C_{32}=12.66$ cGy, $CTDI_{cw32} = 10.55$ cGy), (b) LowZ/XVI ($C_{32}=7.10$ cGy, $CTDI_{cw32} = 5.27$ cGy), (c) 120kVBow/XVI ($C_{32}=1.90$ cGy, $CTDI_{cw32} = 2.97$ cGy).

5. Conclusion and discussion

Cone beam CT images were reconstructed using a LowZ/XVI system and compared with 6MV/iViewGT and kV/XVI systems. Although images with the same CNR as a commercially available CBCT system (100kV/XVI) require excessive doses (>14 cGy) this is significantly less than the circa 240 cGy that would be required for the 6MV/iViewGT system. This highlights the distinct advantages of the LowZ beam over the standard therapy beam for imaging. Although as stated before the LowZ/XVI system cannot produce the same image quality as its kilovoltage counterpart it can produce images with acceptable dose that could be used for the three dimensional positioning of the patient. 3D reconstructions of various anthropomorphic phantoms show that bony anatomy is clearly visible with the LowZ/XVI system using imaging doses equivalent to a few planar MV images from the 6MV/iViewGT system (1-10 cGy).

To further reduce the imaging dose a beam with higher levels of kilovoltage x-rays would be required. In particular a small improvement may be achieved by exploiting coherent bremsstrahlung from a diamond target (T Koenig and U Oelfke 2010). However the biggest improvement would arise if the primary electron beam energy could be significantly reduced. This however would require a linac to operate over a wide range of energies (from very low MeV to 10's MeV) which are not currently available (Connell and Robar 2010).

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