Kilovoltage energy imaging with a radiotherapy linac with a continuously variable energy range.

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Abstract

Purpose:

In this article the effect on image quality of significantly reducing the primary electron energy of a radiotherapy accelerator is investigated using a novel waveguide test piece. The waveguide contains a novel variable coupling device (rotovane) allowing for a wide continuously variable energy range of between 1.4 and 9 MeV suitable for both imaging and therapy.

Method:

Imaging at linac accelerating potentials close to 1 MV was investigated experimentally and via Monte Carlo simulations. An imaging beam line was designed, and planar and cone beam computed tomography images were obtained to enable qualitative and quantitative comparisons with kilovoltage and megavoltage imaging systems. The imaging beam had an electron energy of 1.4 MeV which was incident on a water cooled electron window consisting of stainless steel, a 5 mm Carbon electron absorber and 2.5 mm aluminium filtration. Images were acquired with an amorphous silicon detector sensitive to diagnostic x-ray energies.

Results:

The x-ray beam had an average energy of 220 keV and half value layer of 5.9 mm of copper. Cone

25 beam CT images with the same contrast to noise ratio as a gantry mounted kV imaging system were obtained with doses as low as 2 cGy. This dose is equivalent to a single 6 MV portal image. While 12 times higher than a 100 kVp CBCT system (Elekta XVI), this dose is 140 times lower than a 6MV cone beam imaging system, and 6 times lower than previously published LowZ imaging beams operating at higher (4-5 MeV) energies.

30 **Conclusion**

The novel coupling device provides for a wide range of electron energies that are suitable for kilovoltage quality imaging and therapy. The imaging system provides high contrast images from the therapy portal at low dose, approaching that of gantry mounted kilovoltage x-ray systems. Additionally the system provides low dose imaging directly from the therapy portal potentially
35 allowing for target tracking during radiotherapy treatment. There is the scope with such a tuneable system for further energy reduction and subsequent improvement in image quality.

1 Introduction

Recent advances in radiotherapy have involved improving the conformality of radiation dose to the tumour volume. This has been achieved through improvements in delivery techniques, such as the introduction of intensity modulated radiotherapy (IMRT) and by improved accuracy of patient positioning. The latter advance, commonly known as image guided radiotherapy (IGRT), aims to ensure that the target volume is correctly located with respect to the therapy beam. A variety of techniques can be used to locate and position the target volume.⁶ These systems currently use the therapy beam for imaging, fiducial markers,⁵ gantry mounted kilovoltage imaging systems,¹⁷ in-room kilovoltage systems,³⁴ ultrasound localization,²⁰ radio-frequency markers,² and potentially integrated MRI systems.²⁸

Most of these systems however require add-on systems. Use of the therapy beam for imaging is also advantageous as it allows imaging from the therapy portal, potentially allowing the target volume to be imaged during treatment. However, its use is limited for pre-treatment or inter treatment imaging due to high imaging dose and/or poor image quality (when compared to kilovoltage imaging systems). Potential methods for improving megavoltage imaging have included improving detector quantum efficiency^{24,25,33,39,40} and modification of the megavoltage beam line, such that the linac produces lower energy x-rays more suitable for imaging. The first improvement has resulted in the ability to acquire cone beam computed tomography images using the therapy beam.^{21,35} The second improvement has largely focussed on the use of low atomic number (Z) targets^{3,7-9,11-14,26,29,37} or a combination of medium Z materials and electron absorbers³⁰ to increase the low energy component of the beam.

These systems have shown significant improvements over conventional megavoltage beams with planar imaging doses reduced by a factor of 10 for the same imaging quality.³⁰ CBCT images have

65 also been obtained that are suitable for patient positioning for radiation doses of 1-10 cGy.⁹ However, they remain inferior to kilovoltage gantry mounted imaging systems, requiring imaging doses 50 times higher for the same image quality³¹. Whilst implementing techniques such as coherent bremsstrahlung¹⁹ or improving the panels' quantum efficiency could improve this, the fundamental issue with these systems is the high primary electron energy used to generate the
70 imaging beam. The majority of imaging work has been carried out using an electron beam of around 4 to 6 MeV. Recently linacs operating at energies as low as 2.5 MeV have become available and initial evidence of improved image quality has been reported.³⁶ However, a thorough assessment of the improvement of image quality at these lower electron energies has not been fully explored.

75 Firstly, in this paper characterisation of an Elekta (Elekta, Crawley, UK) waveguide that employs a novel continuously variable coupling device¹ was conducted. This device, hereafter referred to as the rotovane allows for a continuously variable energy range of between 1.4 and 9 MeV suitable for imaging and therapy. An experimental imaging beam line was designed and compared with other
80 radiotherapy x-ray imaging systems. Additionally Monte Carlo models were developed to characterise the system.

2 Materials

2.1 Waveguide test piece

85 The work presented here utilised a waveguide test piece designed by Elekta. The short, 45 cm test piece consisted of an electron gun, buncher section, rotovane, short relativistic section, flight tube and electron window. The electron window was a water cooled, stainless steel construction allowing extraction of a high current electron beam from the vacuum system. RF power was provided by a magnetron and solid state modulator. In this work the effect on beam energy of the

90 novel coupling device (rotovane) was characterised. The rotovane device is an off-axis cell with a
rota-table vane which adjusts the RF coupling between the adjacent on-axis cells ¹. Through
adjustment of the rotavane angle, the amplitude and polarity of the electric field experienced by the
electrons could be finely adjusted in the cells downstream of the rotovane. Crucially for this work it
allowed electrons to be decelerated, thus yielding electrons of lower energy than conventional
95 radiotherapy linacs and which are expected to be useful for producing bremsstrahlung beams
suitable for imaging.

The waveguide operated in a ‘free-running’ mode where no automatic frequency control or ion
chamber feedback was present. For all experiments the waveguide was allowed to reach a steady
100 dose rate and beam current output before data was collected. The dose rate was monitored by a
CC08 ion chamber (IBA, Schwarzenbruck, Germany) and Unidos electrometer (PTW, Freiburg,
Germany PTW).

2.2 Imaging beam line

The imaging beam line consisted of a stainless steel electron window, electron absorber, primary
105 collimator and a secondary collimator system. For the experimental system the secondary
collimation system consisted of lead blocks which were adjusted to desired aperture sizes for
imaging and dosimetry.

Based on our previous work³⁰ the chosen design for the target of the imaging beam consisted of a
thin medium-Z electron window (water cooled stainless steel) to which a low-Z carbon electron
110 absorber was coupled. The thickness of the carbon electron absorber was chosen to be equal to the
practical range of the electrons exiting the electron window. To remove very low energy photons
from the imaging beam 2.5 mm of aluminium was placed on the patient side of the carbon absorber.
This arrangement resulted in an x-ray beam with significantly lower average photon energy than
produced by high-Z target materials.

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Water tanks and imaging phantoms were located at 1000 mm from the source and a detector at 1520 mm. The electron absorber could be removed from the beam to enable measurement of the electron beam characteristics. A photon beam was generated when the absorber was in place.

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2.3 Phantoms

Planar imaging quality was assessed both theoretically and experimentally by utilising the Atlantis phantom previously described.³⁰ In summary, it consists of varying thickness of bone equivalent plastic (from 2 mm to 32 mm in steps of 2 mm) in a tank of water (of 25.8 mm depth) which
125 allowed assessment of contrast and the system signal to noise ratio. CBCT image quality was assessed by utilising a Catphan phantom (The Phantom Laboratory, Salem, USA). The phantom inserts CTP404 and CTP528 were used for assessment of contrast and spatial resolution respectively. Qualitative image quality evaluation for planar and CBCT imaging was undertaken by imaging a anthropomorphic head phantom (RSD, Long beach, USA).

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2.4 Dosimetry equipment

For beam characterisation a 1-D scanning water tank (Type 4322, PTW, Freiburg, Germany) with a thin 3 mm entrance window was used for acquisition of electron and photon depth dose curves. Relative dosimetry was conducted by use of a PPC05 (IBA, Schwarzenbruck, Germany) parallel
135 plate chamber and a CC08 (IBA, Schwarzenbruck, Germany) cylindrical ion chamber. Absolute dosimetry was conducted using a Farmer type chamber (PTW, Freiburg, Germany) and Unidos electrometer (PTW, Freiburg, Germany).

140 **2.5 Monte Carlo model**

BEAMnrc³² and DOSXYZnrc_phsp¹⁸ were used to simulate the imaging beam line and its interaction with the phantoms. The linac model contained all components along the linac beam line, such as the electron window, electron absorbers, primary collimator and secondary collimation. The input electron beam had a nominal radius of 0.5 mm for all simulations
145 irrespective of the input energy spectra. Phase space files were computed at the exit of the linac and subsequently at the detector, after transport through a phantom.

To simulate interaction with the imaging panel a convolution algorithm was used to save calculation time. This involved pre-calculation of the interaction of mono-energetic pencil beams with the imager to determine its response and point spread function as a function of energy. Fifty
150 energy values, evenly distributed on a log10 scale between 0.001 and 10 MeV, were modelled using DOSXYZnrc.³⁸ To calculate the detector signal distribution for a particular beam and phantom, the phase space file was scored at the entrance face of the detector, binned in the fifty energy values, convolved with the response kernels and summed over all energies to yield a predicted image.

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3 Method

3.1 Characterisation of electron energy characteristics of waveguide

Characterisation of the waveguide test piece was conducted by varying the rotovane angle and adjusting the following parameters to obtain maximum beam current:

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- RF peak power – through adjustment of magnetron charge rate and magnetron magnet current.
- RF frequency – adjustment of magnetron frequency to obtain optimal RF frequency in the guide.

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- Gun settings – through adjustment of absolute gun pulse amplitude and gun voltage.

At fixed settings of the RF power and rotovane, the gun settings were adjusted to achieve maximum beam current output. Full characterisation of the waveguide i.e. through adjustment of electron injection delay, differing beam loading settings etc., was beyond the scope of this investigation.

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Optimisation of these parameters could result in lower electron energies and/or improved spot sizes resulting in improved image quality.

The electron beam current for each beam setting was measured using a 50 ohm load placed between the primary collimator and the waveguide, which was held at ground. The electron bunches accelerated by the experimental system were modelled as having a Gaussian energy spread.³⁰

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Characterisation of the energy spread was achieved using the Monte Carlo model of the system. Monte Carlo simulations of the beam line geometry were conducted with a 10x10 cm² field for varying mono-energetic electron beams incident on the vacuum side of the electron window.

Subsequently, depth dose curves were obtained in a water tank with (photon beam) and without

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(electron beam) the 20 mm thick carbon electron absorber and 2.5 mm aluminium in place. Photon

depth dose curves were measured in addition to those of electron beams. This was because at lower MV energies it is difficult to measure electron depth dose curves as the dose maximum lies in or very close to the PMMA side entrance wall of the water tank. A PPC05 chamber and CC08 chamber were used for the electron and photon beams respectively. Electron depth dose curves were used where possible as they are more sensitive to the primary electron energy.

Experimental depth dose curves were obtained using the 1D water tank and PPC05 chamber. The experimental data were then matched to the Monte Carlo depth dose curves using an in-house optimization method. The optimization method took the spectrum of Monte Carlo calculated depth dose curves and weighted them with a Gaussian distribution to find the best match with experiment.

3.2 Experimental characterisation of Imaging Beam

The imaging beam was characterised in using terms of photon depth dose curves and via measurement of the beam half-value layer. The latter was conducted by measuring the relative air kerma (using a Farmer chamber at 100 cm from the source) for varying thicknesses of copper placed on the exit side of the collimator.

Further imaging beam characterisation was obtained using the Monte Carlo model of the system in BEAMnrc. In particular, the average energy of the x-rays emitted by the target was determined using BEAMdp²³ through analysis of a phase space file scored 1 metre beyond the electron window. In addition, the source of photons in the beam line was determined by utilizing the LATCH feature of BEAMnrc to tag where photons had been created.

3.3 Assessment of image quality

Image quality for planar imaging and cone beam computed tomography was determined using the Atlantis and Catphan phantoms respectively. Qualitative image quality was determined from images of anthropomorphic phantoms. Cone beam computed tomography images were obtained by
210 rotating the phantoms on a rotary stage. Comparison of image quality with a megavoltage imaging system (6MV/iViewGT) and a gantry mounted kilovoltage imaging system (Elekta XVI) is also presented based on earlier work^{30,31}. The megavoltage system used a flattened 6MV treatment beam (Elekta, Crawley) and a gadolinium oxysulphide based amorphous silicon panel (iViewGT, Elekta). The XVI system operates at 100 or 120 kVp and employs a columnar CsI based
215 amorphous silicon panel.

3.3.1 Imaging parameters

Imaging of the experimental low energy beam (LowE/XVI) utilised the Elekta XVI columnar caesium iodide based amorphous silicon flat panel imager. The panel was operated in a gated frame
220 read mode whereby the panel was irradiated for 200 ms, and subsequently read out in 142 ms whilst the x-ray beam was gated off. To adjust the dose per frame multiple frames were averaged and/or the pulse repetition frequency was adjusted.

The laboratory measurements were made using the experimental arrangement described in section
225 2.2. The collimation substantially shaped the beam but did not provide the same degree of shielding of a conventional medical linear accelerator. A consequence of this is that some leakage radiation appeared as an additional background in all images. Hence, in addition to being offset and gain corrected the images required removal of the background signal. CBCT reconstruction, including for the XVI system was conducted using an in-house Feldkamp based reconstruction
230 program (Cone.exe, Institute of Cancer Research). All CBCT scans were reconstructed with a

resolution of 1.1 mm, except for the spatial resolution segment of the catphan phantom which was reconstructed at a resolution of 0.55 mm. No scatter correction was performed for any imaging systems.

3.3.2 Planar imaging

Planar image contrast was assessed by using the Atlantis phantom as described in section 2.3. The contrast to noise ratio (CNR) was assessed for varying thicknesses of bone equivalent plastic and for varying dose levels. In this case a water thicknesses of 25.8 cm was used to approximate the thickness of the pelvis region. As the square of CNR is directly proportional to dose¹⁰ a straight line fit was determined to allow the dose for a given CNR to be determined. For the planar images the dose required to achieve the same CNR as the 6MV/iViewGT system was calculated.

3.3.3 ConeBeam CT Contrast to Noise Ratio (CNR)

CNR was assessed by utilising the Catphan phantom. The average CNR between the background region (density = 1.08 g.cm⁻³) and polystyrene and Delrin inserts (density = 1.05 g.cm⁻³ and 1.41 g.cm⁻³ respectively) was determined over 20 slices. The polystyrene and background region were chosen to ascertain the improvement in CNR for materials with small density differences and with attenuation coefficient (μ) differences that do not markedly vary between each other. Whilst this will indicate improvements in soft tissue contrast (for those objects with μ with a small energy dependence) it does not assess, for example, the change in contrast between adipose and soft tissue (whose attenuation coefficients vary markedly below 100 kV). However, this analysis, similar to assessing the low contrast segment of the Catphan for CT image performance¹⁵, provides a relative measure for system comparison. Note that in this case we did not utilise the low contrast segment of the Catphan as it is not possible to see this segment on all imaging beams under comparison.

As with the planar CNR calculation the linear fit parameters were determined for each imaging system and the dose required to match that of the 100 kVp/XVI system was determined. The error associated with the individual points for the straight line fit was determined by finding the error on the mean (95% confidence level) from the 20 slices containing the contrast phantom section.

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3.3.4 Dosimetry

Machine output was measured constantly by a CC08 chamber (IBA) placed on the exit side of the collimator. Output changes were corrected, if necessary (>3%), on a daily basis for dosimetry. Note that the variation of the dose per frame, taken as the variation of the mean pixel value, was less than 1% between frames and 3% over 15 frames. The later variation is due to a general increase in pixel value due to ghosting.

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Planar imaging dose was reported as the dose at the depth of maximum dose (d_{max}) for a 10x10 cm² field for a phantom located at a source to surface distance (SSD) of 95 cm.

CBCT doses were reported as the dose to the centre of either a 16 or 32 cm diameter CTDI phantom (C16 and C32 indices) and via a weighted central slice CTDI index (CTDI_{cw16} or CTDI_{cw32}). The C16 and CTDI_{cw16} indices were quoted for small (<20 cm) phantoms that used a short scan geometry, whilst the C32 and CTDI_{cw32} indices were used for larger phantoms that required a full scan geometry. This method differs from the conventional CTDI index in that it does not average the dose over a 10cm long dosimeter, but instead is calculated on the central portion (slice) of the phantom only. For this study the dose to the centre of the CTDI phantoms was determined via air kerma measurements using a Farmer-type chamber (PTW, Freiburg, Germany).

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Dosimetry for the experimental imaging beam (LowE) was conducted in accordance with TG73.²²

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Note that the beam quality indicator for the LowE system is above the maximum half value layer

(HVL) in the code of practice. Hence, mass attenuation coefficients, backscatter and chamber calibration factors were extrapolated to the required HVL at the required source to surface distance (SSD) and field size. Field sizes were converted from square fields to circular apertures using the summation of rectangles technique.⁴ Dose at depth was determined from Monte Carlo simulations.

4 Results

4.1 Characterisation of waveguide test piece

Figure 1 shows the energy range of the waveguide under differing RF power levels and rotavane positions. Also plotted is a normalised voltage standing wave ratio (VSWR), which is a measure of how efficiently RF power is transmitted into a load. The rotovane position of 0 degrees has been chosen as the mid-point of the VSWR discontinuity, which is associated with the rotovane being in a position in which the vane edge is aligned with the upstream coupling cell hole.

Figure 1 - Electron energy vs. rotavane angle for various input RF power levels. VSWR is Voltage Standing Wave Ratio.

The lowest electron energy achieved was a mean electron energy of 1.4 MeV, determined from matching of experimental and Monte Carlo photon or electron depth dose curves.

4.2 Imaging beam characterisation

The LowE imaging beam line used a carbon electron absorber thickness of 0.5 cm. With this
305 configuration 90% of the photon energy fluence (as measured in a circle of radius 2.5 cm at 100 cm
from the target) is produced in the electron window and the remainder from the carbon absorber.

The imaging beam depth dose curve can be seen in Figure 2, along with other therapy and imaging
systems for comparison. Depth dose curves in all cases are for a 20x20 cm field with an SSD of 95
310 cm. Monte Carlo results also show good agreement with experiment.

Figure 2 - Comparison of depth dose curves for the LowE beam with a kilovoltage imaging system (XVI, Elekta) and a 6MV therapy beam from an Elekta Precise linac. All curves normalised to 100% at a depth of 5 cm.

The measured half value layer of the beam was 5.9 mm of copper. From the Monte Carlo model the average photon energy of the imaging beam was determined to be 220 keV. Additionally the photon spectrum from the imaging beam is shown in Figure 3, and compared to other systems and the XVI detector spectral response.

Figure 3 - Comparison of photon spectra and detector response. LowZ and 6MV spectra obtained from a Monte Carlo model from our earlier work³⁰. SpekCalc is an freely-available program for calculating the x-ray emission spectra from tungsten anode x-ray tubes, based on a published model²⁷

From Figure 3, the LowE beam energy fluence matches the peak detector response to a greater extent than the higher energy megavoltage LowZ system^{30,31} and a 6MV beam. This results in more efficient detection of the photons. Additionally the lower photon energy of the LowE beam results in greater contrast differences in objects with differing mean atomic numbers. This arises due to the photo-electric effect and its Z^3 dependence.

4.2.1 Planar contrast and CNR

The low energy experimental beam (LowE/XVI) shows a factor of two improvement in planar contrast over the standard 6MV/iViewGT system in Figure 4. Good agreement is seen between the experimental LowE/XVI contrast results and Monte Carlo, indicating the Monte Carlo models of the linac, phantom and detector are accurate enough for these purposes.

Figure 4 - Planar contrast for various bone thicknesses in 25.8 cm of water for several imaging beams. Error bars not shown on convolution data as they are too small to show.

Further to the planar contrast results, the dose required to obtain the same CNR as the 6MV/iViewGT system is shown in Table 1. The LowE/XVI system requires only 1.8% of the standard megavoltage image beam dose.

System	Dose (cGy)	% of 6MV Dose
6MV/iViewGT	2	100%
LowE/XVI	0.036	1.8%
120kVp/XVI	0.0020	0.1%

Table 1 – Dose required for the same CNR as a 6MV/iViewGT system (CNR=2.77) for a 1.6 cm bone insert in a 25.8 cm Atlantis phantom.

350 4.2.2 Cone Beam CT Contrast to Noise Ratio

Results are summarised in Table 2. In this table the dose required to obtain the same CNR values as the 100kVp/XVI system are presented. The average standard error on the mean (95% confidence level) of the data points used for the straight line fits was $\pm 20\%$ for the all imaging beams. The LowE/XVI system requires approximately 9 to 12 times more dose than a commercial CBCT system (100 kVp or 120 kVp), but 140 times less than a megavoltage imaging system (6MV/iViewGT). In comparison to megavoltage (4-5 MeV) LowZ imaging systems, the LowE/XVI system requires approximately 6 times less dose for the same CBCT contrast to noise ratio.³¹

355 Catphan images of similar image quality (based on the results in Table 2) are shown in Figure 5.

Beam	Delrin		Polystyrene	
	Dose (cGy)	Dose Ratio to 100kV/XVI	Dose (cGy)	Dose Ratio to 100kV/XVI
100kV/XVI	0.15	1.0	0.15	1.0
120kV/XVI	0.21	1.4	0.18	1.2
LowE/XVI	1.77	11.8	1.42	9.46
LowZ/XVI ³¹	11.00	73.3	7.58	50.5
6MV/iViewGT	244.00	1626.7	n/a	n/a

Table 2 – Dose to give a CNR of 4.9 for the Delrin insert and 2.3 for the polystyrene insert of the Catphan phantom for different imaging systems.

(a)

(b)

Figure 5 - Catphan images for the LowE/XVI and 100kVp/XVI systems. CNR approximately the same in both images. (a) LowE/XVI C16=1.81 cGy and (b) 100 kVp/XVI C16=0.15 cGy.

4.2.3 CBCT spatial resolution

Four line pairs per cm were visible on the Catphan spatial resolution section. This is lower than that achievable on the 100kVp/XVI system of 12 line pairs per cm. Ultimately, spatial resolution in this case is limited by the linac spot size. The spot size on the LowE system could potentially be made smaller if a shorter, non-experimental flight tube and/or focussing bending system were used.

4.2.4 Qualitative image quality

Planar images of a head and neck phantom for the LowE/XVI system are shown in Figure 6 and compared to other systems. Doses were chosen based on the results presented in Table 1, and by qualitative comparison of anatomical structures, such as the vertebra and skull outline. The
 380 LowE/XVI system clearly shows high soft tissue to bone contrast comparable to a kilovoltage imaging system (100kV/XVI), as indicated by visibility of the neck vertebra and skull outline. These structures are present in the 6MV images, but to a lesser extent and for an imaging dose 100 times higher.

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Figure 6 - Images of a head phantom for (a) 6MV/iViewGT (2 cGy)

(b) LowE/XVI (0.015 cGy) and (c) 100kVp/XVI (0.00072 cGy). Images have been histogram equalised.

390 Cone beam computed tomography images of a head phantom are shown in Figure 7 for the LowZ/XVI and 6MV/iViewGT systems. The LowE imaging dose was chosen to produce images with similar contrast to noise ratios based on the results presented in Table 2, whilst maintaining an

image dose acceptable for regular patient positioning (< 10 cGy). Comparison images for the LowZ/XVI and 6MV/iViewGT have been previously published³¹.

(a)

(b)

Figure 7 - 3D reconstructions of a Rando-Alderson head phantom using a short scan geometry for (a) LowE/XVI ($C_{16} = 0.89$ cGy, $CTDI_{cw16} = 0.82$ cGy) and (b) 100kV/XVI normal scan ($C_{16}=0.15$ cGy, $CTDI_{cw16} =0.17$ cGy).

5. Discussion and Conclusion

In this paper an experimental waveguide section, employing a novel coupling device has been characterised, and shown to have a continuously variable energy range between 1.4 and 9 MeV suitable for both imaging and therapy.

Whilst requiring a significant extra dose than a kilovoltage imaging system, kilovoltage equivalent CBCT images of a head phantom were produced with CBCT doses similar to one planar port image (<2 cGy). Image quality was superior to imaging from megavoltage generated LowZ imaging beams.

The tuneable waveguide system evaluated in this work is an experimental system and the lowest currently achievable energy is 1.4 MeV. Optimisation of the waveguide technology and detectors for imaging could include further reduction in the electron beam energy and the use of a higher quantum efficient detector. The Monte Carlo model presented in this paper was used to estimate the potential benefits if the energy could be reduced further to the sub-MV range. Planar contrast was calculated from 0.4 MeV to 6 MeV and the results shown in Figure 8. In addition, experimental results from this work and other systems we have measured³⁰ are presented in this figure. As can be seen the contrast is expected to improve rapidly as the energy is further lowered.

Figure 8 - Planar contrast over a range of beam energies for a 1.6 cm thick bone layer in a 25.8cm thick Atlantis phantom. Beam line geometry was identical to that used for experimental system with the exception of the electron absorber thickness.

The waveguide technology presented here has the potential for producing images with a significant increase in CNR over megavoltage imaging systems, and approaching that of dedicated kilovoltage imaging systems. This technology produces the imaging beam from the therapy beam portal without the need for add-on x-ray systems. The technology opens up the possibility for improved beams eye view tumour tracking by interlacing of the therapy and imaging beams during treatment. It would also be possible to use such a system in conjunction with a kilovoltage imaging system to perform three dimensional tracking during VMAT or for increased speed for a standard cone beam CBCT acquisition¹⁶.

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