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FULL PAPER

Investigation of the use of external aluminium targets for portal imaging in a medical accelerator using Geant4 Monte Carlo simulation

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Objective: To install a low-Z target on the wedge tray mount of a medical linear accelerator to create a new image beam and to confirm image contrast enhancement.

Methods: Experimental low-energy photon beams were produced with the linac running in the 6 MeV electron mode and with a low-Z target installed on the wedge tray mount [denoted 6 MeV (low-Z target)]. Geant4 Monte Carlo simulation was performed to analyse the energy spectrum and image contrast of a 6 MeV (low-Z target) beam. This study modelled the 6 MeV (low-Z target) beam and the 6 MV (megavoltage) radiotherapy photon beam and verified model validity by measurement. In addition, a contrast phantom was modelled to quantitatively compare the image contrasts of the 6 MeV (low-Z target) beam and the 6 MV radiotherapy photon beam. A low-Z target was fabricated to generate low-energy photons (25–150 keV) from incident electrons, and a portal image of the Alderson RANDO phantom was acquired using a clinical linear accelerator for qualitative analysis.

Results: The measured and calculated percentage depth dose of the 6 MV photon and 6 MeV (Al) beams were consistent within 1.5 and 1.6%, respectively, and calculated lateral profiles of the 6 MV photon beam and the 6

MeV (Al) beam were consistent with the measured results within 1.5 and 1.9%, respectively. Although low-energy photons (25–150 keV) of the 6 MV photon beam were only 0.3%, the Be, C, and Al low-Z targets, but not the Ti target, generated 34.4 to 38.5% low-energy photons. In 5 to 20 cm water phantoms, contrast of the 6 MeV (Al) beam was approximately 1.16 times greater than that of the 6 MV beam. The contrasts of 6 MeV (Al) and 6 MV photon beams in the 20 cm water phantom were ~34% lower than those in the 5 cm water phantom. 6 MeV (Al)/CR (computed radiography) images of the human body phantom were more vivid and detailed than 6 MV/EPID (electronic portal imaging device) and 6 MeV (Al)/EPID images.

Conclusion: The experimental beam with a low-Z target, which was simply installed on the wedge tray mount of the radiotherapy linear accelerator, generated significantly more low-energy photons than the 6 MV radiotherapy photon beam, and provided better quality portal images.

Advances in knowledge: This study shows that, unlike the existing low-Z beam studies, a low-Z target can be installed outside the head of a linear accelerator to improve portal image quality.

INTRODUCTION

Over the last two decades, the development of image-guidance systems for external radiation therapy has resulted in continual improvements in patient positioning and target localization. For these systems, the megavoltage (MV) portal imager and the kilovoltage (kV) on-board imager (OBI) are mainly used to identify the accuracy of patient

setup. The MV portal imager has the advantage of short image acquisition times, but suffer from poor image quality mainly due to the Compton effect.¹ On the other hand, the kV-OBI system provides good visualization of the target in two-dimensional or three-dimensional with an additional high-resolution soft tissue information.^{2,3} However, the latter has some drawbacks, such as, geometric mismatch

between treatment and image isocentre, image artefacts from high-Z materials, greater costs for maintenance, increased quality assurance and lack of beam's eye viewing.^{4,5} For these reasons, the kV-OBI system cannot completely replace the MV portal imaging system which is invariably used when OBI system fail. Thus, efforts should be made to improve the MV portal image quality which requires a large proportion of diagnostic photons in the energy range 25–150 keV.^{6,7} For that purpose, the high atomic number target and flattening filter should be removed from the linear accelerator head to reduce the self-absorption of low-energy photons.

Many research groups have shown that the use of low atomic number (Z) targets can improve image quality as compared with standard MV therapy beams.^{1,4,6–16} In the majority of the previous studies, low-energy photon beams were generated by modifying the head configuration of the medical linear accelerator (linac) to install a low-Z target or by modifying the conventional treatment beam. However, these methods of beam generation are complex and require the assistance of the manufacturer. In this study, we aimed to generate a low-energy photon beam that improves image contrast by simply installing a low-Z target on a medical linear accelerator. Monte Carlo simulations (MCSs) of linear accelerators from several manufacturers have been widely used in low-Z beam studies.^{1,8,9,12,16} When the correct input parameters are used, an IAEA (International Atomic Energy Agency) format phase space file can produce accurate dose distributions that compare well with measurements.^{17–19} In this study, we modelled low-Z target beam generation by using the IAEA phase space file as the beam source, and fabricated a low-Z target to confirm image contrast enhancement.

The purposes of this study were to analyse the effects of two beam generation parameters (target atomic number and thickness) and the phantom thickness by installing a low-Z target on a wedge tray mount, and characterize image quality improvements. A Geant4 MCS^{20–22} was performed to quantify the effects of various parameters, and the results were verified experimentally. A portal image of the Alderson RANDO phantom was obtained using a low-Z generated photon beam for the qualitative analysis of image quality, and this was also compared with images obtained using a 6 MV beam.

METHODS AND MATERIALS

Generation of 25–150 keV photon beams

In previous low-Z portal imaging studies, several low-Z targets were installed in the empty ports of the linac carousel.^{1,7,12,13} In the present study, a low-Z target was installed on the linac wedge tray mount to improve portal imaging contrast. Experimental low-energy photon beams were produced with the linac running in the 6 MeV electron mode and with a low-Z target installed on the wedge tray mount [denoted 6 MeV (low-Z target)]. As in previous studies, a target material of low atomic number was used to generate a photon beam in the diagnostic energy range (25–150 keV). The target materials were beryllium (Be, Z = 4), carbon (C, Z = 6), aluminium (Al, Z = 13), and titanium (Ti, Z = 22). Interactions between a 6 MeV electron beams with slabs ranging from 4 to 24 mm were modelled. A portal imaging

Table 1. Parameters for Geant4 Monte Carlo simulation

Electron beam source (6 MeV)	IAEA format phase space data (myvarian.com/montecarlo)	73.3 cm from isocentre
Jaw field size	20 × 20 cm ²	At isocentre
X, Y jaw	Tungsten (15 g cm ⁻³)	Y 68.1 cm from isocentre X 59.4 cm from isocentre
Mylar window	MYLAR (1.4 g cm ⁻³)	45 cm from isocentre
Polyethylene filter	Polyethylene (0.94 g cm ⁻³)	41 cm from isocentre
Low-Z target	Aluminium (2.699 g cm ⁻³) Beryllium (1.848 g cm ⁻³) Carbon (2.000 g cm ⁻³) Titanium (4.54 g cm ⁻³)	40 cm from isocentre

experiment was performed to identify the target material and target thickness that most efficiently generated low-energy photons (25–150 keV) from incident electrons.

Geant4 Monte Carlo simulation

Low-Z target generated beams and 6 MV photon beams modelling

A Geant4 (v. 9.6 patch03) MCS with an option 3 physics list was used to for photon energy spectral analysis and to quantify image contrast. The cut-off ranges of photons, electrons, and positrons were all 0.01 mm. The low-Z target beam generated from 6 MeV electron beams and 6 MV photon beams were modelled. Table 1 and Figure 1 provide the geometry and a schematic diagram of the linear accelerator used for Geant4 simulation. Phase space files provided by Varian were used as the beam source for 6 MV photons and 6 MeV (low-Z target) beams.^{10,17,19} IAEA format phase space files and related data are available from the manufacturer (at myvarian.com/montecarlo). The IAEA format phase space file was generated from a Geant4 Monte Carlo code using the geometric input parameters of a clinical linear accelerator. Energies and spectral distributions of incident electrons were tuned to fit measured dose distributions. The IAEA format phase space was recorded above the upper jaws. Thus, the phase spaces of 6 MV photons and 6 MeV electrons were used as the beam source by placing them on the upper jaws. The upper jaw, the lower jaws, the Mylar window, the low-Z targets, and the polystyrene filters were modelled and simulated in Geant4 code. The size, shape, and materials used for jaws and the window were in accord with an engineering drawing provided by the manufacturer. The upper and lower jaws were set at a field size of 20 × 20 cm² at the isocenter. Several low-Z target materials and polystyrene filters were installed onto the wedge tray mount in slab form.

Spectral data acquisition

Slabs of beryllium (Be, Z = 4), carbon (C, Z = 6), aluminium (Al, Z = 13) and titanium (Ti, Z = 22) were used as targets. As shown in Figure 1, the low-Z target was located on the wedge

Figure 1. Schematic diagram of the medical linear accelerator head components used to generate 25–150 keV photon beams.

tray mount of the accelerator (Clinac iX), and the scoring plane was located immediately below the target. The energy spectra of particles generated from targets of different thickness were recorded and analysed. The spectra of low-Z target beams were compared with that of the 6 MV photon beam. The histories of incident photons and electrons for 6 MV photon beams and 6 MeV (low-Z target) beams were 2.47×10^9 and 7.04×10^9 , respectively, and thus, statistical uncertainty was within 2%.

Contrast phantom modelling, contrast data acquisition

The contrast phantom simulated in Geant4 was similar to that used by Parsons et al⁹. A rectangular parallelepiped ICRP bone (3 cm long, 3 cm wide, and 5 cm high) surrounded by water was located at the centre of the isocentre. The 9×9 cm water phantom was simulated by varying phantom thickness (5, 10, 15, and 20 cm) to determine the relation between phantom thickness and changes in contrast, which was calculated after recording energy fluences by placing a phase space (10 cm deep from isocentre) under the contrast phantom. Contrast was calculated using:

$$\text{contrast} = \frac{\Psi_{\text{bone}} - \Psi_{\text{water}}}{\Psi_{\text{water}}}$$

where Ψ_{bone} and Ψ_{water} are the average energy fluences recorded in the bone and water regions of the phase space.

EXPERIMENTS

Aluminium target, polystyrene filter

A photograph of a low-Z target and an electron filter is shown in Figure 2. The target material was aluminium (Al, $Z = 13$) and

Figure 2. The custom made low-Z target and polystyrene filter.

6 MeV incident electron beams were used to produce a low-energy photon beams. Multiple aluminium targets with thicknesses of 5 or 1 mm were fabricated. Target thicknesses were not sufficient to stop all incident electrons, and thus, transmitted electrons were absorbed by the polystyrene filter, which was placed immediately below the aluminium target. The ratios between photons generated in the low-Z target and incident electrons were analysed by changing electron filter thickness from 4 to 20 mm by MCS. This simulation was used to determine the thickness of the optimal polystyrene electron filter, and this was then applied to generate aluminium target beams.

Validation of beam modelling

To verify the accuracy of 6 MeV (Al) and 6 MV photon beam modelling, dose distributions calculated by using Geant4 were compared with measured data. Percentage depth dose (PDD) and lateral profiles were obtained by Blue phantom (IBA Dosimetry, Germany) and a Farmer type ion chamber (FC65-G, IBA Dosimetry, Germany). PDD was measured at a source surface distance (SSD) of 100 cm for a field size of 20×20 cm². Lateral profile was measured at maximum dose depth (d_{max}) and a SSD of 100 cm for a field size of 20×20 cm². A dose rate of 1000 MUm⁻¹ was used to measure the 6 MeV (Al) beam. Due to the relatively low beam current in electron mode, doses were cumulatively recorded for 1 min at each measurement point; this method enabled the detector to obtain a relatively high signal. To obtain dose distributions by MCS, a water phantom was constructed using the same set up used in practice. The overall size of the water phantom was $50 \times 50 \times 50$ cm, and the voxel size was $5 \times 5 \times 5$ mm. In the water phantom, the cut-off ranges of photons, electrons and positrons were all 0.1 mm.

Rando phantom portal image acquisition

The detector used for image acquisition from the 6 MV photon beam was an electronic portal imaging device (EPID). A typical EPID contains a 1.0 mm copper conversion plate to improve the detection efficiency of the radiotherapy megavoltage spectrum. When imaging using the photon beam generated with a low-Z target, the copper conversion plate significantly attenuates photons within the diagnostic range, and thus, reduces image quality.¹⁴ Computed radiography (CR) was used for the 6 MeV (Al) beam because the beam contained a low-energy photon beam. The portal images of 6 MV and 6 MeV (Al) beams were acquired by placing the head and neck region of the Alderson RANDO Phantom (The Phantom Laboratory, Salem, NY) at the isocentre. Source to imager distances of the EPID and CR were 150 cm and 110 cm, respectively. 6 MV beam images were acquired for EPID and CR and compared with CR images of 6 MeV (Al) beams.

Figure 3. Percentage depth dose curves obtained by measurement and modelling for 6 MV and 6 MeV(Al) beams for a field size of $20 \times 20 \text{ cm}^2$ at a source-to-surface distance of 100 cm.

RESULTS

Figure 3 shows PDD curves of a 6 MV photon beam and a 6 MeV (Al) beam for a field size of $20 \times 20 \text{ cm}^2$ at a SSD distance of 100 cm. Each PDD curve was normalized with respect to maximum dose. The statistical uncertainty of doses calculated by Monte Carlo simulation was $<2\%$. Monte Carlo simulation results of depth dose distribution for the 6 MV photon and 6 MeV (Al) beams were consistent with experiment data within 1.5 and 1.6%, respectively. Parsons et al showed that 98.3% of calculated points for the 6 MV flattened beam with a field size of $10 \times 10 \text{ cm}^2$ were consistent within 1% at a depth of $>0.85 \text{ cm}$.⁹ Constantin et al reported 95% of calculated points for a 6 MV flattened beam for a field size of $10 \times 10 \text{ cm}^2$ were consistent within 1% at all depths.¹⁹ The maximum dose depths (d_{max}) of the 6 MV photon beam and the 6 MeV (Al) beam were 15 mm and 12.5 mm, respectively. As was expected, the PDD curve of 6 MV photons exhibited beam hardening as compared with that of 6 MeV (Al) beams. Percentage depth dose at depth of 10 cm (PDD_{10}) were 70% for 6 MV and 63% for 6 MeV (Al).

Figure 4. Cross-line profiles of measured and modelled 6 MV and 6 MeV (Al) beams at a depth of 1.5 cm, a field size of $20 \times 20 \text{ cm}^2$, and a source-to-surface distance of 100 cm.

Figure 5. Calculated low-energy (25–150 keV) photon spectra for 6 MeV electron beams using different target materials and for 6 MV photon beam.

Figure 4 shows the lateral profile curves of a 6 MV photon beam and a 6 MeV (Al) beam for a field size of $20 \times 20 \text{ cm}^2$ at a SSD of 100 cm and a depth of 1.5 cm. All profile curves were normalized with respect to dose values on the central axis. The maximum difference between measured data and results calculated by MC simulation for 6 MV photon beam and 6 MeV (Al) beam lateral dose distributions were 1.5 and 1.9%, respectively, and the positions of 50% dose were consistent within 1 and 1.6 mm respectively. Constantin et al showed that the maximum dose percentage difference of the lateral profile for a 6 MV photon beam up to a field size of $30 \times 30 \text{ cm}^2$ was 2% within the field.¹⁹ Because the experimental low-energy photon beams are forward peaked beam profile in the absence of a flattening filter it was necessary to take background flood field correction to enable contrast calculation.

Figure 5 shows the spectral distributions of various low-Z target materials for the 6 MeV (low-Z target) beam. These distributions were obtained at target thicknesses with the greatest proportions of low-energy photons (25–150 keV). Although other results are not shown in the figure, simulations were performed for targets of various thicknesses, and the spectrum of the 6 MV photon beam is also shown for comparison purposes. For each spectrum, the fractional fluence for low-energy photons (25–150 keV) is indicated, and typically increased on decreasing atomic number. Although the low-energy photon beam of the 6 MV photon beam was only 0.3%, the low-Z beams of Be, C, and Al targets (but not Ti) contained 34.4 to 38.5% low-energy photon beams. This result was similar to the 25 to 150 keV photon ratio of the energy spectrum reported in a previous study, in which the low-Z target material was inserted at a different position.⁸

Figure 6a shows the ratio of low-energy photons generated per incident electron for Be, C, Al, and Ti targets of different thicknesses. The photon generation efficiency curves of targets were normalized with respect to maximum values. For comparison purposes, C, Al, and Ti results were normalized by 0.85, 0.65, and 0.45, respectively. Be, C, Al, and Ti targets showed maximum

Figure 6. Relative efficiencies based on fluences exiting targets for low-Z targets (a) with different thicknesses, and (b) different atomic numbers.

photon generation efficiencies at thicknesses of 18, 14, 8, and 4 mm, respectively. Figure 6b shows the relation between low-energy photon generation efficiency and target atomic number, which was used to identify the best target material. Again for comparison purposes, efficiencies were normalized with respect to the efficiency of Al, which was slightly higher than that of carbon, but approximately 11 and 18% greater than those of beryllium and titanium, respectively. Table 2 compares the ratios of photons generated by 6 MeV (Al) beams and incident electrons at various polystyrene electron filters thicknesses. The ratio of remaining

incident electrons decreased on increasing polystyrene electron filter thickness. The results obtained showed that at 20 mm polystyrene electron filter eliminated nearly all incident electrons without changing the low-energy photon ratio.

Figure 7 shows the energy fluence map of the cortical bone generated from using 5, 10, 15, and 20 cm thick water phantoms using the 6 MeV (Al) beam and the 6 MV photon beam. Energy fluence was obtained when the phase space was located at a source axis distance (SAD) of 110 cm. All images were normalized to a flood field image, which was obtained with the contrast phantom removed. This was done to account the forward nature of the low-Z beams and the characteristic of the 6 MV photon beams not completely flat. In addition, each energy fluence image was normalized with respect to the average energy fluence of water, which was obtained from a 5 cm water phantom and a 6 MeV (Al) beam. The contrast of the 6 MeV (Al) beam was greater than 6 MV photon beam at all phantom thicknesses. Contrast of the 6 MeV (Al) beam was approximately 1.16 times greater than that of the 6 MV beam. Furthermore, contrast decreased on increasing phantom thickness for 6 MeV (Al) and 6 MV beams. The contrasts of 6 MeV (Al) and 6 MV photon beams in the 20 cm water phantom were ~34% lower than those in the 5 cm water phantom. The result means that the advantages of the 6 MeV (Al) over the 6 MV photon beam is diminished in imaging of thicker objects because of the greater attenuation of the low-energy component of the spectrum in a thicker object. This issue reveals the need for further investigation in greater patient thicknesses.

Rando phantom portal images were acquired to determine how low-Z beams affected patient images. Anteroposterior and lateral images of the head and neck region obtained using EPID and CR for 6 MeV (Al) and 6 MV photon beams are shown in Figure 8; images had no window level correction. As was expected, image contrast significantly increased using a CR that could detect low-energy photons. Furthermore, 6 MeV (Al)/CR images described the human body phantom in more vivid and detailed manners than 6 MV/EPID and 6 MeV (Al)/EPID images. Contrast was relatively good and more clearly differentiated bone, air, soft tissue and other tissues. In addition, fine parts of teeth, jawbone, and the cervical region were clearly visible.

DISCUSSION

MCSs of several linear accelerators have been used to study low-Z target beams. The IAEA format phase space file provided by the manufacturer does not require a complicated accelerator head or input parameter optimization, and thus, enables efficient MCS. It has further been reported that the application of a phase space file provides more accurate dose distributions than

Table 2. Comparison between low-energy (25–150 keV) photons and electrons generated by 6 MeV using an aluminium target for different polystyrene electron filter thicknesses

Polystyrene thickness (mm)		None	4	6	8	10	12	14	16	18	20
Proportion (%)	All Photons	64.2	73.9	79.7	85.6	90.7	94.7	97.3	98.6	99.3	99.5
	Low E photons (25–150 keV)	34.4	34.6	34.7	34.7	34.8	34.8	34.8	34.7	34.7	34.7
	Electrons	35.8	26.1	20.3	14.4	9.3	5.3	2.7	1.4	0.8	0.5

Figure 7. Images of energy fluences in the phase space for 5, 10, 15, 20 cm thick water phantoms generated with low-Z and 6 MV photon beams. All images were obtained same greyscale level.

measurements.^{17–19} However, phase space data are not useful for design studies of low-Z targets located in the medical linac head because phase space data record particle information immediately above the jaws. In the present study, a Geant4 Monte Carlo model was developed using an IAEA format phase space file for the spectral analysis of low-Z targets located on a wedge tray mount. In addition, a standard 6 MV photon beam and a low-Z beam were modelled by applying minimum head structures that were not included when recording the IAEA phase space file. Calculated dose distributions of the low-Z and 6 MV photon beams were consistent with measurements, indicating beam modelling was relatively accurate.

Several authors have calculated the spectra of low-Z targets located in the accelerator head. The spectrum of the 6 MeV (Al) beam obtained in this study was similar to those of the 4 MeV/Al (Clinac) beam published by Orton and Robar⁸ and of the 2.35 MV/Carbon (TrumBeam) beam published by Parsons *et al*.⁹ The ratio of low-energy photons for the standard 6 MV treatment beam was 0.3%, and peak fluence occurred at ~500 keV. These results are consistent with the spectral results of a standard 6 MV treatment beam published by Orton and Robar⁸. In agreement with that previously reported for a low-Z target beam in the linear accelerator head, the low-Z target beam located on the wedge tray mount was found to significantly increase the ratio of low-energy photons as compared with the standard 6 MV beam.

Flampouri *et al* reported that a 4 MeV electron beam showed maximum efficiency when a 6 mm thick Al target was used.¹³ Orton and Robar used a 1 cm thick Al target.⁸ In the present study, the thickness of the optimal Al target was 8 mm (Figure 6), and the thickness of the Al target we used for the experimental beam corresponds to approximately 60% of the continuously slowing down approximation range for the aluminium, which is similar to the optimal Al target thickness found in the previous study.

Figures 7 and 8 show that the image quality of the 6 MeV (Al) beam was better quantitatively and qualitatively than the standard 6 MV beam. Several research groups have evaluated image quality by modelling EPID.^{8,9,16} Instead of modelling the image detector, we analysed image contrast using energy fluence of the phase space. The current study was a preliminary study of image improvements achievable using a low-Z target beam installed on the wedge tray mount. Image quality was not evaluated in more detail and the detector was not optimized. These issues will be the subjects of future studies.

The use of low-Z targets and low-energy electron beams can improve image quality as compared with MV treatment beams. Compared with the kV imaging system, the low-Z beam has the advantages of low mechanical complexity, cost reduction, and low QA requirements. In this respect, many authors have studied low-Z targets located on linac heads to increase the ratio of low-energy photons in the diagnostic energy range. In the present study, we developed and verified a simple means of improving image quality using a low-Z target.

CONCLUSION

This study modelled a low-Z target beam installed on a wedge tray mount using Geant4 Monte Carlo code. The results of the MCS were verified by comparing the calculated dose distributions of 6 MV photon and 6 MeV (Al) beams with measured data. PDD and lateral profile were consistent with measured values. Significant spectral differences between the low-Z beam and the standard 6 MV treatment beam were observed. The low-Z beam generated a significantly higher amount of lower energy photons in the diagnostic energy range than the 6 MV photon beam. Furthermore, contrast phantom modelling showed that the 6 MeV (Al) beam had better contrast than the 6 MV photon beam. In portal image experiments using the Rando phantom, image contrast was improved when detectors capable of detecting low-energy photons were used.

Figure 8. Images of the head and neck region of a Rando anthropomorphic phantom, (a) 6 MV photon with EPID (b) 6 MV photon with CR (c) 6MeV electron/Al target with CR. All images were obtained using same window level. EPID, electronic portal imaging device.

In conclusion, this study confirms that beams generated by a low-Z target installed on the wedge tray mount produce images of higher quality than the standard 6 MV photon beam.

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