A Monte Carlo investigation of low-Z target image quality generated in a linear accelerator using Varian’s VirtuaLinac\textsuperscript{a)}

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Purpose: The focus of this work was the demonstration and validation of VirtuaLinac with clinical photon beams and to investigate the implementation of low-Z targets in a TrueBeam linear accelerator (Linac) using Monte Carlo modeling.

Methods: VirtuaLinac, a cloud based web application utilizing Geant4 Monte Carlo code, was used to model the Linac treatment head components. Particles were propagated through the lower portion of the treatment head using BEAMnrc. Dose distributions and spectral distributions were calculated using DOSXYZnrc and BEAMdp, respectively. For validation, 6 MV flattened and flattening filter free (FFF) photon beams were generated and compared to measurement for square fields, 10 and 40 cm wide and at $d_{\text{max}}$ for diagonal profiles. Two low-Z targets were investigated: a 2.35 MeV carbon target and the proposed 2.50 MeV commercial imaging target for the TrueBeam platform. A 2.35 MeV carbon target was also simulated in a 2100EX Clinac using BEAMnrc. Contrast simulations were made by scoring the dose in the phosphor layer of an IDU20 aSi detector after propagating through a 4 or 20 cm thick phantom composed of water and ICRP bone.

Results: Measured and modeled depth dose curves for 6 MV flattened and FFF beams agree within 1% for 98.3% of points at depths greater than 0.85 cm. Ninety three percent or greater of points analyzed for the diagonal profiles had a gamma value less than one for the criteria of 1.5 mm and 1.5%. The two low-Z target photon spectra produced in TrueBeam are harder than that from the carbon target in the Clinac. Percent dose at depth 10 cm is greater by 3.6% and 8.9%; the fraction of photons in the diagnostic energy range (25–150 keV) is lower by 10% and 28%; and contrasts are lower by factors of 1.1 and 1.4 (4 cm thick phantom) and 1.03 and 1.4 (20 cm thick phantom), for the TrueBeam 2.35 MV/carbon and commercial imaging beams, respectively.

Conclusions: VirtuaLinac is a promising new tool for Monte Carlo modeling of novel target designs. A significant spectral difference is observed between the low-Z target beam on the Clinac platform and the proposed imaging beam line on TrueBeam, with the former providing greater diagnostic energy content. © 2014 Author(s). All article content, except where otherwise noted, is licensed under a Creative Commons Attribution 3.0 Unported License. [http://dx.doi.org/10.1118/1.4861818]

Key words: low-Z target, flattening filter free, Monte Carlo

1. INTRODUCTION

Monte Carlo simulation of low atomic number (Z) targets in conventional linear accelerators (Linacs), such as the Varian Clinac series\textsuperscript{1–6} (Varian Medical Systems, Inc., Palo Alto, CA), Elekta SL25 and Precise Treatment Systems\textsuperscript{7–9} (Elekta Oncology Systems, Crawley, UK) or Siemens Primus and Oncor Linacs\textsuperscript{10–12} (Siemens Medical Solutions, Concord, CA), have all used geometric and material specifications of the Linac provided by the manufacturer to help construct accurate models in BEAMnrc.\textsuperscript{13} However, this technical information for Varian’s TrueBeam Linac platform (Varian Medical Systems, Inc., Palo Alto, CA) has not been made available to the research community for proprietary reasons. Currently, International Atomic Energy Agency (IAEA) format phase space files for individual clinical photon beams (4, 6, 8, 10, and 15 MV, plus 10 and 15 MV flattening filter free) are available from the manufacturer (at myvarian.com/montecarlo), generated using the Geant4 Monte Carlo code\textsuperscript{14} and geometry input from computer aided designs.\textsuperscript{15,16} While the phase space files have been shown to produce accurate dose distributions compared to measurement,\textsuperscript{17} they are not useful for investigating novel target designs in TrueBeam since they correspond to a location just above the secondary collimation (jaws). However, Varian has developed a cloud based web application, called “VirtuaLinac,” which allows for Monte Carlo simulation without releasing the full description of the Linac in this section. In this work, VirtuaLinac is used to simulate both the...
6 MV flattened beam and the 6 MV flattening filter free (FFF) photon beam for validation purposes, and the results compared to measured dose distributions. Next, a carbon target operated at 2.35 MeV and a newly proposed 2.50 MeV commercial imaging target beam (Varian Medical Systems, Inc., Palo Alto, CA) were then modeled. The low-Z target beams were compared to a previously validated \(^1\) Monte Carlo simulation of a carbon target operated at 2.35 MeV generated in a Varian 2100EX Clinac using BEAMnrc.

2. MATERIALS AND METHODS

2.A. Varian’s VirtuaLinac

Photon beam generation in the TrueBeam Linac platform was simulated using Varian’s VirtuaLinac application. VirtuaLinac is a cloud based web application running on Amazon Web Services (aws.amazon.com, Inc., Seattle, WA) that utilizes the Geant4 Monte Carlo code to model TrueBeam. Multiple instances can be created to run simulations in parallel. Compared to individual institutions each implementing their own model, advantages of VirtuaLinac are that (i) one can start running simulations using already implemented geometry and (ii) multiple users of the same code enable a stronger validation. VirtuaLinac allows the user to change parameters of the model such as incident electron energy, energy spread, spot size, position, and angular divergence, as well as target composition, thickness, location, and the choice of filtering filter. Proprietary information including clinical target design, filtering filter design, and the location and composition of components within the Linac head are hidden from the user. The simulation can be used to produce phase space files in planar geometry at a customizable distance from isocenter. We recorded the phase space immediately upstream of the jaws, 73.3 cm from isocenter. The phase space file was streamed to a local computer. The format of the phase space files is compatible with the IAEA specification, \(^18\) intended to facilitate import into different Monte Carlo codes.

2.B. Monte Carlo photon beam simulations

In this work, eight-eight-core computers were used to simulate the treatment head above the jaws. Version 9.4.patch2 of Geant4 was utilized, with the option 3 physics list including Rayleigh scattering. A range cut of 75 \(\mu\)m was used for photons, electrons, and positrons, where the range cut distance is internally converted to energy for individual materials within the model. This serves as the energy threshold for the production of secondary particles. Four photon beams were generated: the 6 MV (nominal) flattened beam, the 6 MV FFF beam, the proposed 2.50 MeV commercial imaging beam (denoted 2.5 MV/IB) and for comparison to previous work \(^5\) a carbon target operated at 2.35 MeV (denoted 2.35 MV/carbon). The 6 MV flattened photon beam was modeled with incident electron energy of 6.18 MeV, an energy spread of 0.053 MeV. For the 6 MV FFF, an incident electron energy of 5.90 MeV was used, with an energy spread of 0.051 MeV. For the 2.5 MV/IB, an energy spread of 0.3 MeV was used. As in previous work \(^5\) the carbon target beams were generated with a 2.35 MeV mono-energetic electron beam. As described by Sawkey et al. \(^19\), the lateral profiles of the spot size were taken to be Gaussians, with sigmas of 0.6866 and 0.7615 mm (6 MV flattened), 0.6645 and 0.7274 mm (6 MV FFF), 0.72 and 0.76 mm (low-Z beams) in the crossplane and inplane directions, respectively. Spot sizes were determined by measurements on 6 MV flattened and FFF beams. The electron beam was incident normally on the target, with no angular divergence. The incident energies and spectral distributions were the values used to generate the phase space files available on myvarian.com/montecarlo. The treatment head models that generated these phase spaces were initially built by Varian Medical Systems. The energy spectrum of the incident electron beams were tuned to match dose distributions measured by Chang et al. \(^20\) Phase space files with this tuning were recorded immediately above the jaws are available myvarian.com/montecarlo. The 6 MV, 6 MV FFF, and 2.5 MV/IB target designs were implemented according to engineering drawings and not disclosed to the user. The target for the 2.35 MV/carbon beamline was a 6.7 mm thick disc of carbon (\(\rho = 2.00 \text{ g/cm}^3\)) with a diameter of 5 cm. Target thickness was modified from previous work \(^2\) to account for the change in material density specified according to the values in Geant4. The flattening filter and the brass plate were removed for both the 2.35 MV/carbon and 2.5 MV/IB beams. Phase space files of \(4.8 \times 10^7\) particles for a \(10 \times 10 \text{ cm}^2\) field (at isocenter) were created, with an uncertainty of approximately 1% in measured dose at \(d_{\max}\) for a 6 MV flattened and FFF beam. These required \(1.8 \times 10^{10}, 1.5 \times 10^{10}, 3.1 \times 10^{10}, \text{ and } 5.1 \times 10^{10}\) incident electron histories for the 6 MV, 6 MV FFF, 2.50 MV/IB, and 2.35 MV/carbon photon beams, respectively.

Upon completion of the simulations in VirtuaLinac, the IAEA phase space files were propagated through 3 mm of air in BEAMnrc and summed using BEAMdp. \(^21\) The result was one EGSnrc format phase space file for each photon beam located 8.9 mm above the y-jaws. These phase space files were then propagated through the lower portion of the treatment head using BEAMnrc for a range of field sizes defined by the jaws. This process was used as a result of difficulties in summing the IAEA phase space files with BEAMdp. For clinical photon beams, global electron (ECUT) and photon (PCUT) cut-off energies of 0.700 and 0.010 MeV, respectively, were used. Similarly, AE = 700 MeV and AP = 0.01 MeV, where AE and AP are the low energy thresholds for the production of knock-on electrons and secondary bremsstrahlung photons, respectively. For low-Z target beams, AE = ECUT and AP = PCUT values of 0.521 and 0.01 MeV, respectively, were used.

For comparison with previous work \(^5\), the 2.35 MV/carbon photon beam was simulated in a Varian 2100EX Clinac using BEAMnrc. The 2.35 MV/carbon model was modified from previous work \(^5\) to account for changes present in the most recent High Energy Accelerator Monte Carlo Data Package provided by Varian Medical Systems under a non-disclosure agreement. These modifications largely consisted of the inclusion of the metal plating on the dielectric windows of...
the monitor chamber. $5.0 \times 10^7$ incident electron histories were run for the 2.35 MV/carbon photon beam. Directional bremsstrahlung splitting was used with a splitting radius of 10 cm for a $10 \times 10$ cm², at a source-to-surface distance (SSD) of 100 cm, with a bremsstrahlung splitting number of 2000. AE = ECUT and AP = PCUT values of 0.521 and 0.010 MeV, respectively, were used. BEAMdp was used to determine spectral distributions of the phase space files at isocenter, for a $10 \times 10$ cm² field. For visual comparison, the resulting spectral distributions were normalized by the area under the curve.

2.C. Validation of TrueBeam photon beams

Percentage depth dose (PDD) measurements of the 6 MV flattened and FFF clinical photon beams were recorded for $10 \times 10$ cm², in water, at a SSD of 90 cm. Diagonal profiles were measured at $d_{max}$, for a field size of $40 \times 40$ cm², with an SSD of 90 cm. Measurements were acquired using a $50 \times 50 \times 50$ cm³ water tank (Scanditronix, Uppsala, Sweden) and a 0.125 cm³ cylindrical ion chamber (PTW N31010, Freiburg, Germany). A dose rate of 1400 MU/min was used for measurement of FFF beams, as the ion collection efficiency is only decreased by 0.4% compared to using a dose rate of 600 MU/min. Corresponding Monte Carlo dose distributions were run using DOSXYZnrc. Lateral voxel dimensions of 2 and 4 mm were used, within the central region and penumbra, respectively. Vertical voxel dimensions of 2, 5, and 10 mm was used along central axis for depths between 0 and 4, 4 and 9.5, and 9.5 and 35.5 cm, respectively. AE = ECUT and AP = PCUT were set to 0.521 and 0.010 MeV, respectively. Dose distributions of the 2.5 MV/IB and 2.35 MV/carbon beams in TrueBeam and Clinac were simulated in DOSXYZnrc with the same water phantom for a $10 \times 10$ cm² field, at a SSD of 100 cm.

2.D. Monte Carlo model of the IDU20

Similar to the methods used by Orton and Robar and Connell and Robar, an IDU20 detector was modeled using DOSXYZnrc. Previously, Munro and Boutis have shown that 99.5% of the detector response is a result of dose to the phosphor screen, therefore dose was scored in the phosphor layer to estimate image contrast. The copper layer in the IDU20 was removed for the imaging photon beams, as the presence of the plate has been previously shown to reduce image contrast-to-noise ratio by a factor ranging from 1.4 to 3.2 for a 2.35 MV/carbon beam generated within a 2100EX Clinac. The contrast phantom simulated in BEAMnrc consisted of a 4 cm long cylinder of ICRP bone with a diameter of 3 cm centered on isocenter and surrounded by water. Additionally, for representation of a more typical patient separation, each side of the above contrast phantom was padded with 8 cm of water, for a total thickness of 20 cm. The IDU20 was placed 150 cm from the source. For comparison to the imaging photon beams, contrast for the 6 MV flattened and FFF photon beams were simulated with the copper layer in place. Contrast was calculated in MATLAB (Mathworks, Natick, MA) as

$$\text{contrast} = \frac{|P_{\text{Bone}} - P_{\text{Water}}|}{P_{\text{Water}}}$$

where $P_{\text{Bone}}$ and $P_{\text{Water}}$ are the average scored dose in the bone and water regions in the phosphor. To account for the forward nature of the low-Z target beams, the images were normalized with flood fields taken with an expanded field and the contrast phantom removed.

3. RESULTS AND DISCUSSION

3.A. Validation of 6 MV flattened and FFF photon beams

Figure 1 shows PDD curves for measured and modeled data for the 6 MV flattened and FFF beams, for a field size of $10 \times 10$ cm², at a SSD of 90 cm. Figure 3(a) shows the corresponding difference in percent dose between the measured and modeled data for depths ranging from 0.85 to 30 cm. 98.3% of values at depths greater than 0.85 cm show agreement within 1% or less for both the 6 MV flattened and FFF beams. For a $40 \times 40$ cm² field (not shown), 94.0% and 95.7% of values at depths greater than 0.85 cm show agreement within 1% or less for the 6 MV flattened and FFF beams, respectively. For comparison, Constantin et al. have shown a 1% agreement for 95% of values in depths ranging from 5 to 30 cm for a 10 × 10 cm², 6 MV flattened field, with 98% of calculated points agreeing within 2% between 2 and 40 cm. Gете et al. have reported less than 1% agreement for all depths greater than 0.5 cm for a 6 MV FFF $10 \times 10$ cm² field, using cylindrical phase spaces generated by Constantin et al.

Figure 2 shows diagonal profiles for measured and modeled data for the 6 MV flattened and FFF beams measured at $d_{max}$, for a $40 \times 40$ cm² square field, at a SSD of 90 cm. Figure 3(b) shows the corresponding gamma factor analysis
for measured and modeled diagonal profiles. For the criteria of 1.5 mm and 1.5%, 93.1% and 95.6% of points analyzed for a 6 MV flattened and FFF beam, respectively, had a gamma value less than one, with the source of gamma values greater than one occurring in the penumbra region. Constantin et al.\textsuperscript{16} have shown a similar agreement for 6 MV flattened photon beam, with lateral profiles at a depth 10 cm showing a maximum percent dose difference of 2% and 1% in-field and out-of-field, respectively, for field sizes up to 30 × 30 cm\(^2\). At larger depths (20 cm), they report a maximum difference of 4% for a 30 × 30 cm\(^2\) field, with a 2% agreement for smaller fields. Similarly Gete et al.\textsuperscript{17} have reported an agreement better than 1% for 95% of calculated points, at depths greater than 1.5 cm for a 6 MV FFF, for field sizes greater than 3 × 3 cm\(^2\). Modeled diagonal profiles were approximately 0.7 mm narrower than the measured profiles at 50% dose for both the 6 MV flattened and FFF beams. Similarly, Gete et al.\textsuperscript{17} have reported a difference between modeled and measured 6 MV FFF photon beam at 50% dose ranging from 1.0 to 1.4 mm for lateral profiles of a 40 × 40 cm\(^2\) field, at a depth of 10 cm, and a SSD of 100 cm. Similar to the results of Constantin et al.,\textsuperscript{16} we have an unresolved discrepancy of approximately 4% in the beam horns between the measured and modeled diagonal profiles. While the discrepancy is slightly different from that reported by Constantin et al.,\textsuperscript{16} it is unlikely caused by an error in the mean incident electron energy or the FWHM of the electron spot size, as these parameters were tuned to match measured in-plane and cross-plane dose profiles. The cause of this discrepancy is currently unresolved for the 6 MV flattened beam diagonal profile.

### 3.B. Simulation of low-Z target photon beams

Figure 4 shows relative spectral distributions for the 2.5 MV/IB and the 2.35 MV/carbon photon beams in TrueBeam (produced using Geant4) and Clinac (produced using EGSnrc) between 10 and 150 keV. The low-Z target beam energy spectra in TrueBeam are appreciably harder than that produced in a Clinac. This is observed in the decrease in the relative fraction of diagnostic energy photons (25–150 keV) by 10% and 28% for the 2.35 MV/carbon and 2.50 MV/IB beams, respectively, in TrueBeam, compared to the 2.35 MV/carbon beam produced in Clinac. Compared to previous work, the 2.35 MV/carbon spectrum in TrueBeam more closely resembles a 4 MV/aluminum target beam in Clinac, reported by Orton and Robar,\textsuperscript{2} with a diagnostic energy population of 37%, with no change in the diagnostic energy.

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Fig. 2. Diagonal profiles of measured and modeled values for 6 MV flattened and flattening filter free photon beams, at \(d_{\text{max}}\), for a 40 × 40 cm\(^2\) square field, at a source-to-surface distance of 90 cm.

Fig. 3. (a) Percentage dose difference between the measured and modeled data points for 6 MV flattened and flattening filter free photon beams, for a 10 × 10 cm\(^2\) square field, at a source-to-surface distance of 90 cm. (b) Gamma factor analysis of measured and modeled diagonal profiles for 6 MV flattened and flattening filter free beams, at \(d_{\text{max}}\), for a 40 × 40 cm\(^2\) square field, at a source-to-surface distance of 90 cm.

Fig. 4. Relative spectral distributions for a 2.35 MV/carbon target in TrueBeam and Clinac as well as a proposed commercial 2.50 MV/IB in TrueBeam at isocenter, for a 10 × 10 cm\(^2\) field. Shown are the fractions of photons in the diagnostic energy domain (25–150 keV).
FIG. 5. Percentage depth dose curves for 2.35 MV/carbon target beam produced in TrueBeam and Clinac, with a field size of 10 × 10 cm² at a source-to-surface distance of 100 cm. The energy percentage produced TrueBeam when using their energy range. Furthermore it is considerably harder than the previously reported 2.35 MV/aluminum target beam produced in a Clinac, with a diagnostic energy population of 46%. The 66 keV spectral peak of the 2.35 MV/carbon beam in TrueBeam is closer to that of the 4 MeV titanium target beam produced in Clinac, reported by Tsechanski et al. with a peak of 80 keV. Similarly, the 2.5 MV/IB spectrum is similar to the 4 MeV copper target beam produced in Clinac, reported by Tsechanski et al., with a spectral peak at approximately 120 keV. The 2.35 MV/carbon beam produced in Clinac, with a peak energy fluence at 70 keV, is comparable to that reported by Roberts et al., with a peak energy fluence at approximately 75 keV. This beam was produced in an Elekta Linac with a 1.9 MeV electron beam incident on a thin stainless steel electron window coupled to a carbon electron absorber and 2.5 mm aluminum filter.

Figure 5 shows PDD curves for the 2.5 MV/IB and 2.35 MV/carbon beams produced in TrueBeam and Clinac, with a field size of 10 × 10 cm² at a SSD of 100 cm. The PDD curves also indicate beam hardening in TrueBeam compared to Clinac. The percent dose at a depth of 10 cm (PDD10) was 53.9% for the 2.5 MV/IB, 48.6% and 45.0% for the 2.35 MV/carbon target beams produced in TrueBeam and Clinac, respectively. This represents an increase in PDD10 of 3.6% and 8.9% for the 2.35 MV/carbon and 2.5 MV/IB target beams in TrueBeam, respectively, compared to the 2.35 MV/carbon target beam in Clinac. PDD10 for the 2.35 MV/carbon beam produced in TrueBeam is similar to the 2.35 MV/aluminum (PDD10 = 50.8%) beam produced in Clinac, with a difference of 2.2%.

Figure 6 shows dose distributions in the phosphor layer in the IDU20 for the 4 and 20 cm thick imaging phantoms in the beam line for the 2.35 MV/carbon beams in Clinac and TrueBeam, 2.5 MV/IB, 6 MV flattened, and 6 MV FFF photon beams. Image contrast was normalized to the contrast of the dose distribution produced by 2.35 MV/carbon beam generated in Clinac. The 4 cm thick phantom clearly illustrates the effect of the harder spectra produced in TrueBeam, with a decrease in contrast by a factor of 1.1 and 1.4 for the 2.35 MV/carbon and 2.5 MV/IB beams produced in TrueBeam, respectively, compared to the 2.35 MV/carbon beam produced in Clinac. This effect is diminished between the two 2.35 MV/carbon beams for the 20 cm thick phantom, with only a reduction in contrast by a factor 1.03. This decrease is largely due to the attenuation of low energy photons within the 2.35 MV/carbon beam produced in Clinac.

FIG. 6. Images the dose distribution in the phosphor layer of an IDU20 detector of the thin (top row) and thick (bottom row) phantoms with the copper layer in the IDU20 removed for the low-Z target beams.
thereby not contributing to image formation. However, a significant difference remains between the 2.35 MV/carbon and 2.50 MV/IB beams, with a decrease in contrast by a factor of 1.4 and 1.3, compared to the 2.35 MV/carbon beam produced in Clinac and TrueBeam, respectively. It is notable that a significant portion of the 2.35 MV/carbon spectrum produced in TrueBeam aligns well with the previously reported peak detector response of the IDU20, of approximately 60 keV, while only a minimal amount of these energy photons existing in the 2.50 MV/IB spectrum. Only a minimal difference is observed between the two phantom thicknesses for the 6 MV flattened and FFF photon beams. However, factor differences in contrast of 4.3 and 3.3 (4 cm thick phantom) and 3.7 and 2.7 (20 cm thick phantom) remain for the 6 MV flattened and FFF beams compared to the 2.35 MV/carbon beam produced in Clinac. Given the limited availability of the TrueBeam imaging target at the time of writing we cannot present measured results in the present work; however, this will be the subject of future investigation.

4. CONCLUSIONS

In this work, we have shown that VirtuaLinac is a useful tool for Monte Carlo modeling of clinical and novel low-Z target photon beams. The Monte Carlo simulations were validated by comparing the 6 MV flattened and FFF modeled dose distributions to measured data, with PDD agreements within 1% or less for greater than 98.3% of points at depths deeper than 0.85 cm. When low-Z targets are simulated in TrueBeam, the spectra were hardened compared to that produced in a Clinac for a 2.35 MV/carbon beam, with a decrease in the diagnostic photon energy population. This resulted in a reduction in image contrast for the 2.35 MV/carbon and 2.50 MV/IB beams produced in TrueBeam, compared to the 2.35 MV/carbon beam produced in Clinac. In conclusion, a significant spectral difference is observed between the low-Z target beam on the Clinac platform and the proposed imaging beam line on TrueBeam, with the former providing greater diagnostic energy content.

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*VirtuaLinac is available by contacting one of the authors at montecarloresearch@varian.com or through the website RadiotherapyResearchTools.com.

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J. L. Robar, “Generation and modelling of megavoltage photon beams for contrast-enhanced radiation therapy,” Phys. Med. Biol. 51(21), 5487–5504 (2006).

E. J. Orton and J. L. Robar, “Megavoltage image contrast with low-atomic number target materials and amorphous silicon electronic portal imagers,” Phys. Med. Biol. 54(5), 1275–1289 (2009).

T. Connell and J. L. Robar, “Low-Z target optimization for spatial resolution improvement in megavoltage imaging,” Med. Phys. 37(1), 124–131 (2010).

A. Tsechanski, A. F. Bielajew, S. Faermann, and Y. Krutman, “A thin target approach for portal imaging in medical accelerators,” Phys. Med. Biol. 43(8), 2221–2236 (1998).

D. Parsons and J. L. Robar, “Beam generation and planar imaging at energies below 2.40 MeV with carbon and aluminum linear accelerator targets,” Med. Phys. 39(7), 4568–4578 (2012).

D. Parsons and J. L. Robar, “The effect of copper conversion plates on low-Z target image quality,” Med. Phys. 39(9), 5362–5371 (2012).

S. Flamourou, P. M. Evans, F. Verhaegen, A. E. Nahum, E. Spezi, and M. Partridge, “Optimization of accelerator target and detector for portal imaging using Monte Carlo simulation and experiment,” Phys. Med. Biol. 47(18), 3331–3349 (2002).

D. A. Roberts, V. N. Hansen, A. C. Niven, M. G. Thompson, J. Seco, and P. M. Evans, “A low Z linac and flat panel imager: Comparison with the conventional imaging approach,” Phys. Med. Biol. 53(22), 6305–6319 (2008).

D. A. Roberts et al., “Kilovoltage energy imaging with a radiotherapy linac with a continuously variable energy range,” Med. Phys. 39(3), 1218–1226 (2012).

D. Z. Ostapiak, P. F. O’Brien, and B. A. Faddegon, “Megavoltage imaging with low Z targets: Implementation and characterization of an investigational system,” Med. Phys. 25(10), 1910–1918 (1998).

B. A. Faddegon, V. Wu, J. Pouliot, B. Gangadharan, and A. Bani-Hashemi, “Low dose megavoltage cone beam computed tomography with an unflattened 4 MV beam from a carbon target,” Med. Phys. 35(12), 5777–5786 (2008).

D. Sawkey et al., “A diamond target for megavoltage cone-beam CT,” Med. Phys. 37(3), 1246–1253 (2010).

D. W. Rogers, B. A. Faddegon, G. X. Ding, C. M. Ma, J. We, and T. R. Mackie, “BEAM: A Monte Carlo code to simulate radiotherapy treatment units,” Med. Phys. 22(5), 503–524 (1995).

S. Agostinelli et al., “GEANT4: A Simulation toolkit,” Nucl. Instrum. Meth. A506(3), 250–303 (2003).

M. Constantin, D. E. Constantin, P. J. Keall, A. Narula, M. Svatos, and J. Perl, “Linking computer-aided design (CAD) to Geant4-based Monte Carlo simulations for precise implementation of complex treatment head geometries,” Phys. Med. Biol. 55(8), N211–N220 (2010).

M. Constantin et al., “Modeling the TrueBeam linac using a CAD to Geant4 geometry implementation: Dose and IAEA-compliant phase space calculations,” Med. Phys. 38(7), 4018–4024 (2011).

E. Gete et al., “A Monte Carlo approach to validation of FFF VMAT treatment plans for the TrueBeam linac,” Med. Phys. 40(2), 021707 (13pp.) (2013).

INDC International Nuclear Data Committee, Phase-Space Database for External Beam Radiotherapy Summary Report of a Consultants’ Meeting (International Atomic Energy Agency, Vienna, Austria, 2006).

D. Sawkey et al., “Measurement of incident electron spots on TrueBeam,” Med. Phys. 40(6), 332 (2013).

Z. Chang et al., “Commissioning and dosimetric characteristics of TrueBeam system: Composite data of three TrueBeam machines,” Med. Phys. 39(11), 6981–7018 (2012).

C. M. Ma and D. W. O. Rogers, BEAMdp Users Manual NRCC Report PIRS-0509/CirrevA (NRCC, Ottawa, Canada, 2009).

S. Lang, J. Hrbacek, A. Leong, and S. Klock, “Ion-recombination correction for different ionization chambers in high dose rate flattening-filter-free photon beams,” Phys. Med. Biol. 57(9), 2819–2827 (2012).

I. Kawrakow, “Accurate condensed history Monte Carlo simulation of electron transport. I. EGStm, the new EG54 version,” Med. Phys. 27(3), 485–498 (2000).

H. Munro and D. C. Bouius, “X-ray quantum limited portal imaging using amorphous silicon flat-panel arrays,” Med. Phys. 25(5), 689–702 (1998).