Polish Journal of Medical Physics and Engineering
The Journal of Polish Society of Medical Physics
ISSN 1898-0309, doi: 10.2478/pjmpe-2025-0012

June 2025
Vol 31, Issue 2

Scientific Paper

Equivalent spectral and filtration models of the Halcyon 2.0 kV-CBCT system

Jonathan Mbewe [1,ABCD,*]

[1]*Division of Medical Physics, Groote Schuur Hospital/University of Cape Town, Cape Town, South Africa*
**Corresponding author: Jonathan Mbewe; jmbewe@gmail.com*

(received 9 August 2024; revised 10 January 2025; accepted 17 April 2025)

Abstract

Introduction: Monte Carlo simulations are the gold standard for radiation dosimetry. However, developing accurate models for kilovoltage cone-beam computed tomography (kV-CBCT) systems can be challenging because accurate spectral and geometry specifications from vendors are not always available. This study presents equivalent spectral and filtration models that may be used for Monte Carlo modeling of the kV-CBCT beamline on-board the Halcyon 2.0 linear accelerator.

Material and methods: Equivalent energy spectra for the 100, 125, and 140 kV beams were determined by matching beam half-value layers and air kerma outputs calculated using SPEKTR 3.0 software with measurements. The bowtie filter profile was reconstructed from transmission measurements, enabling the generation of a 3D filter model. The model was incorporated into Geant4/GATE Monte Carlo simulations for the purpose of computing depth and off-axis dose characteristics. Calculated percentage depth doses (PDDs) and off-axis profiles were benchmarked against those measured in a water phantom.

Results: The maximum differences between measured and computed PDDs were 3.47% (100 kV), 3.65% (125 kV), and 3.27% (140 kV). For off-axis profiles, maximum differences were 3.73% (100 kV), 6.84% (125 kV), and 4.44% (140 kV) within the central beam. Larger discrepancies up to 22% occurred in high-gradient penumbral regions due to mismatches in spatial resolution between detectors and Monte Carlo scoring geometry.

Conclusions: This study presents the first Monte Carlo model of the Halcyon 2.0 kV-CBCT system using measurement-derived equivalent models. The agreement between computed and experimental dose characteristics validates the accuracy of the model for imaging dose calculations until vendor supplied data is made accessible for geometry and spectral specifications.

Keywords: Monte Carlo, imaging dose, kV-CBCT, GATE

Introduction

Kilovoltage cone-beam computed tomography (kV-CBCT) plays a vital role in image-guided radiotherapy (IGRT), ensuring accurate patient positioning and reducing interfractional uncertainties. In addition, kV-CBCT coupled with modern reconstruction algorithms generate high quality images thereby enabling adaptive radiotherapy[1]. However, the additional radiation exposure from kV-CBCT can contribute to cumulative doses that exceed Organs-At-Risk (OAR) thresholds and potentially increase the risk of secondary solid tumours. To mitigate this, the American Association of Physicists in Medicine (AAPM) Task Group 180 recommends that imaging doses remain within 5% of the prescribed therapeutic dose; any excess must be accounted for in treatment planning[2]. Accurate quantification of kV-CBCT dose is therefore essential for optimization and risk assessment.

Monte Carlo simulations are the gold standard for radiation dosimetry due to their ability to accurately model radiation transport and complex geometries[3]. However, developing precise Monte Carlo models for kV-CBCT systems is challenging, as crucial information on source characteristics and system geometry - essential for accurately modelling photon fluence and attenuation - is often proprietary and only available under Non-Disclosure Agreements, if at all.

To address this limitation, researchers have developed semi-empirical methods to estimate source and filtration characteristics. Turner et al. proposed a technique that derives equivalent energy spectra and filtration models from measurements of output, Half-Value Layers (HVLs), and bowtie filter attenuation using an ionisation chamber on multidetector CT scanners[4]. Their models were used in Monte Carlo simulations for dosimetric calculations and generated results deemed superior

Authors' contribution: A – Research concept and design, B – Collection and/or assembly of data, C – Data analysis and interpretation, D – Writing the article, E – Critical revision of the article, F – Final approval of the article.

to those generated with vendor-supplied data as they accounted for age-related deterioration in performance. McMillan et al. later applied this approach to a kV-CBCT system on board a radiotherapy linear accelerator[5]. Li et al. further refined the method by using a linear detector array to profile the bowtie filter, significantly reducing the number of required measurements while generating bowtie filter profiles at each kV setting of the scanner[6].

In this study, we applied the methodologies of Turner et al. and McMillan et al.[4,5] to model the kV-CBCT beamline of the Halcyon 2.0 linear accelerator (Varian Medical Systems, Palo Alto, CA, USA). Since vendor-provided specifications, including energy fluence spectra and bowtie filter geometry, are not publicly available for this system, our approach offers a practical alternative for beam modelling.

Despite the growing adoption of the Halcyon, the lack of publicly available manufacturer data presents challenges for accurate quantification of imaging dose through Monte Carlo simulations. Existing studies have focused on measurements of point doses in anthropomorphic phantoms using TLDs[7] or simply measurement of protocol-specific dose indices[8]. However, only modelling permits a more accurate determination of 3D dose distributions in reference computational phantoms or even patient-specific computational phantoms. This study addresses this gap by developing and validating equivalent source and filtration models of the Halcyon 2.0 kV-CBCT. These models may be used in Monte Carlo simulations for dose assessments. To our knowledge, this is the first work to provide such models, offering a practical alternative for kV-CBCT imaging dose estimation on the Halcyon.

Materials and methods

Halcyon kV-CBCT system

The Halcyon 2.0 linear accelerator is a ring-gantry radiotherapy system designed to streamline access to advanced Image-Guided Radiotherapy (IGRT) while reducing the complexity of commissioning such programs. It integrates a 6 Megavoltage (MV) inline linear accelerator, imaging devices, and other peripherals within a compact ring-gantry design. To simplify implementation, the system comes with preconfigured machine models, beam-source models, radiation dose calculation algorithms for therapeutic dose modelling, and CBCT imaging protocols.

The first-generation Halcyon utilized Megavoltage Cone-Beam Computed Tomography (MV-CBCT) for imaging, using the therapy beam. Later versions introduced a kilovoltage (kV) CBCT system, featuring a dedicated kV x-ray source equipped with a fixed half-bowtie filter. All kV-CBCT protocols employ full-gantry rotation, offering fields of view (FOV) between 281 mm and 491 mm. The x-ray source operates at peak voltages from 40 kV to 150 kV, with a focal spot positioned 100 cm from the isocentre.

Imaging is performed using a 43 cm × 43 cm amorphous silicon detector with a resolution of 1028 × 1028 pixels, providing a scan FOV of 49.1 cm. The imager is laterally offset by 17.5 cm from the isocentre, maintaining a constant Source-to-Imager Distance (SID) of 154 cm. The ring-gantry architecture enables faster imaging, achieving a maximum rotation speed of 4 rotations per minute (RPM), significantly outpacing conventional C-arm LINACs, which typically reach only 1 RPM. **Figure 1** illustrates the spatial configuration of the kV and MV sources alongside their respective imagers.

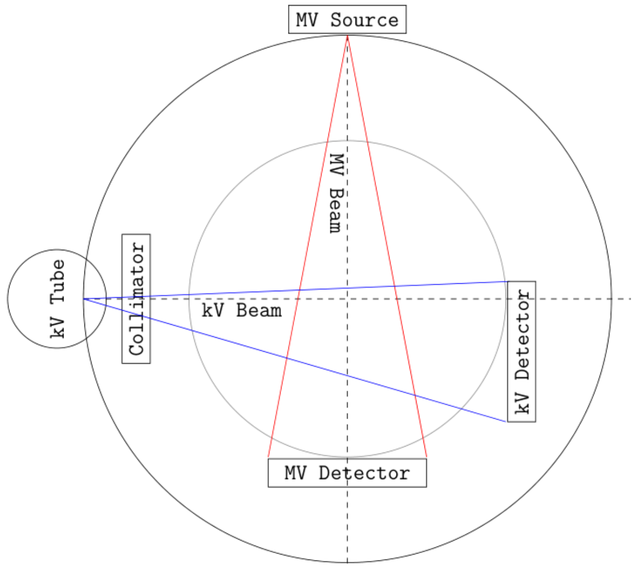

Figure 1. Illustration of the kV and MV beamline arrangements on the Halcyon 2.0.

Characterisation of the Halcyon kV source

The detailed specifications of the Halcyon kV-CBCT x-ray source, including energy-fluence spectra, bowtie filter geometry, and other attenuation characteristics, are not publicly available. To overcome this limitation, we derived equivalent spectral and filtration models using measurement-based techniques inspired by Turner et al.[4], originally developed for multi-detector CT, and later adapted for kV-CBCT by McMillan et al.[5]. This approach requires no prior knowledge of the x-ray tube's spectral or filtration properties, aside from the assumption that the spectra originate from a Tungsten target.

Determination of equivalent energy spectra

The equivalent energy spectra represent the calculated x-ray spectra that most accurately replicate the characteristics of the actual x-ray tube based on measurement data. Specifically, these spectra are determined by ensuring that their first and second Half-Value Layers (HVL1 and HVL2) and x-ray tube output (measured in mGy/mAs) closely match the corresponding experimental measurements.

Measurement of HVL1, HVL2, and tube output

The broad-beam geometry method was used to determine the first and second Half-Value Layers (HVL1 and HVL2) of the beam. Dosimetric measurements were performed using a calibrated solid-state XR detector and MagicMax module (IBA Dosimetry GmbH, Schwarzenbruck, Germany). Two

30 cm-high Styrofoam blocks were placed on either side of the detector to support variable-thickness aluminium absorbers (99.9% pure).

For each x-ray tube voltage setting (100, 125, and 140 kV), the tube current and exposure time were set to 100 mA and 100 ms, respectively. Five exposures were taken both with and without filters, and the average reading was calculated. The process was repeated while progressively adding aluminium filters until the average air kerma dropped just below one-quarter of the initial no-filter measurement. The resulting air kerma vs. absorber thickness curve was used to determine HVL1 and HVL2. The experimental setup is illustrated in **Figure 2**.

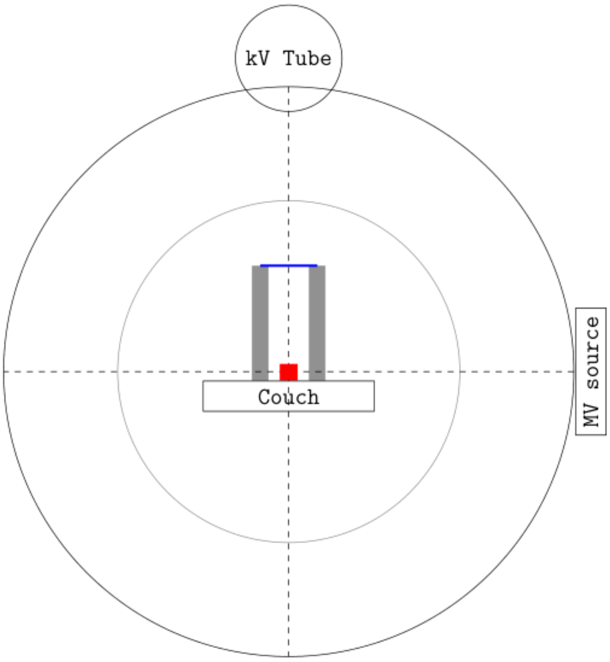

Figure 2. Experimental setup for HVL measurements at the Halcyon. The red square represents the detector position, the grey vertical rectangles represent the Styrofoam blocks, while the blue horizontal immediately above the Styrofoam blocks represents the Aluminium filter position.

Additionally, tube current was measured using the MagicMax current probe, which was invasively connected to the x-ray generator during exposures without aluminium filtration.

Calculation of equivalent energy spectra

To calculate the equivalent energy spectra, we used SPEKTR (v3.0), a computational tool for x-ray spectrum modelling and analysis[9]. Written in MATLAB, SPEKTR employs the Tungsten Anode Spectral Model with Interpolating Cubic Splines (TASMICS), formulated by Hernandez and Boone, to simulate spectral and exposure characteristics of generic tungsten x-ray tubes[10]. The software generates unfiltered energy spectra at a 1.0 keV resolution based on input parameters such as peak tube voltage (kVp) and % ripple. A built-in function, spektrTuner, optimises the initial spectra by adjusting them to match measured and theoretical x-ray tube output (in mGy/mAs).

Using SPEKTR, we generated energy spectra for the three kV settings and fine-tuned them to minimise discrepancies between measured and calculated tube output. Initially, an equivalent bowtie filter composed of aluminium was assumed. Two additional absorbers were incorporated: (1) a hardening material representing the cumulative inherent filtration and (2) an aluminium filter with a thickness equal to the measured HVL2 for each kV setting. The thickness and material composition of the inherent filtration were iteratively adjusted until the calculated HVL2 matched the measured values.

The final spectrum was considered representative of the x-ray beam transmitted through the central portion of the bowtie filter, assumed to be 0.5 mm thick. Since the Halcyon's bowtie filter is fixed and non-removable, we performed an additional recalculation in SPEKTR—this time excluding the 0.5 mm bowtie filter - to estimate the spectrum emitted directly from the x-ray source. The resulting spectra were saved as two-column text files: the first column contained energy bins at 1 keV resolution up to 150 keV (x-values), while the second column listed the corresponding photon fluence (Photons/mm^2/mAs) at isocentre (y-values).

Measurement of the bowtie profile

The equivalent bowtie filter's varying thicknesses were determined through transmission measurements of the actual, but unknown, half-bowtie filter in the Halcyon's kV x-ray system. To achieve this, the gantry was positioned at 180°, placing the x-ray tube at the 3 O'clock position. A PTW 30013 Farmer-type ionisation chamber (PTW Freiburg, Freiburg im Breisgau, Germany) was securely mounted at the superior edge of the treatment couch in air (**Figure 3**). The chamber's stem was stabilised on a small rectangular piece of Prestik affixed to the couch top, while a piece of masking tape ensured the detector remained immobilized.

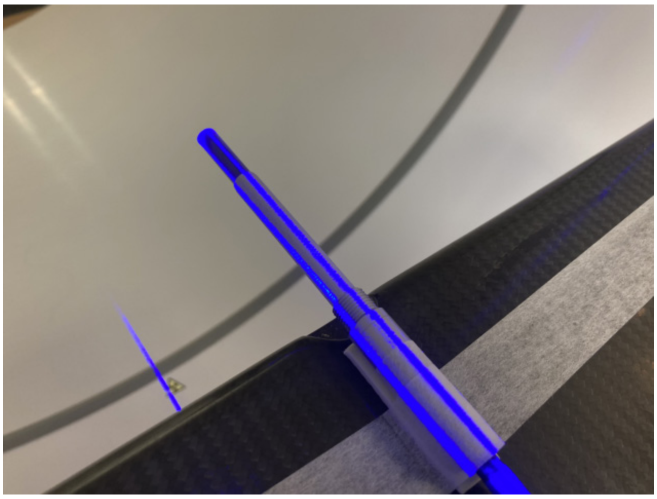

Figure 3. Placement of a Farmer type ionisation chamber for profiling the bowtie shape on the Halcyon kV-CBCT system.

To map the bowtie filter's transmission profile, exposures were taken starting from the anterior edge of the imager. The ionisation chamber was then incrementally moved in 0.5 to 1.0 cm steps across the bowtie filter until it reached the posterior edge of the imager. The measurement setup is illustrated in **Figure 4**.

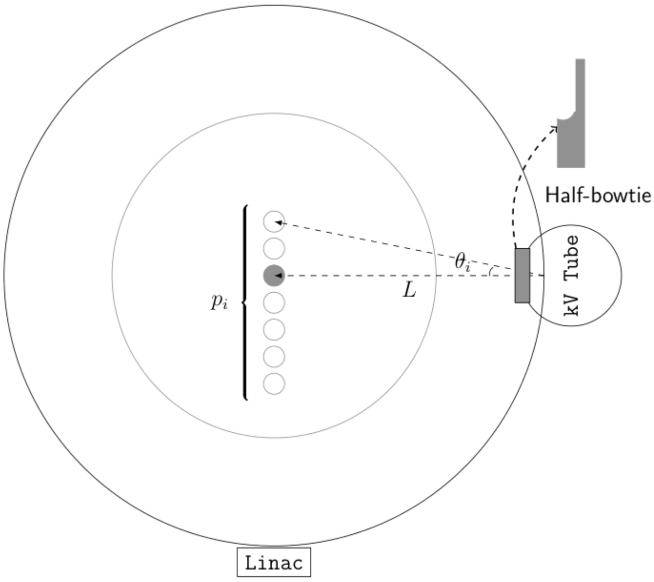

Figure 4. Experimental setup for bowtie profile measurement at the Halcyon. The circles inside the gantry bore represent ion chamber positions (pi) along the vertical axis, forming angles ϕi with the Source Axis Distance (L) on central beam axis at the x-ray tube focus. The distance pi is measured from the kV imaging isocentre.

The angle ϕ_i corresponding to each measurement position was determined using the focal spot-to-isocentre distance L and the vertical displacement of the ionisation chamber, p_i, as given by Equation (1):

$$\phi_i = \tan^{-1}(p_i/L) \quad (1)$$

The charge measured at each position p_i (corresponding to angle ϕ_i) was expressed as a fraction of the charge measured at p_0 (where $\phi_0 = 0$, i.e., at the isocentre). To construct an equivalent bowtie filter, the following steps were performed:

1. Using SPEKTR, an initial 0.5 mm thick aluminium filter was added to represent the central portion of the bowtie filter, t_o. The air KERMA at isocentre was then calculated using the equivalent spectra.
2. An additional aluminium filter of thickness t_1 was introduced, representing the effective path length (L_{BT}) through the bowtie filter at angle ϕ_1. The air KERMA was recalculated, and t_1 was iteratively adjusted until the ratio of calculated air KERMA at p_1 and p_0 matched the corresponding ratio of measured charges at those positions.
3. Steps 1 and 2 were repeated for all ionisation chamber positions p_i, generating a bowtie filter profile by plotting p_i against t_i.

To determine the filter's surface coordinates, we adopted the method proposed by Zhang et al.[11]. **Figure 4** illustrates the bowtie filter's spatial location in the XY (axial) plane with the focal spot as the origin. The curved surface coordinates (x, y) of the bowtie filter were then calculated using Equations (2a) and (2b):

$$x(\varphi) = y_1 \cdot \tan\varphi - L_{BT}(\varphi) \cdot \sin\varphi \quad (2a)$$

$$y(\varphi) = y_1 - L_{BT}(\varphi) \cdot \cos\varphi \quad (2b)$$

Since the bowtie filter was assumed to have a flat inline profile, no profiling was performed in that direction.

Generating the 3D model of the bowtie filter

The (x, y) coordinates of the bowtie filter's curved section for each angle ϕ were determined using Equation (2), with y_1 = 147 mm - based on the dimensions of an x-ray tube used in the kV-CBCT system on the Varian TrueBeam linear accelerator - and t_0 = 0.5 mm, as assumed in the equivalent bowtie filter model.

These coordinates were then used to plot discrete points in the Computer-Aided Design (CAD) software FreeCAD[12]. A Cartesian coordinate system was established, placing the tube focal spot at the origin. A cubic spline was fitted through the plotted points to create the curved section of the filter, while straight-line segments were added to represent its flat surfaces.

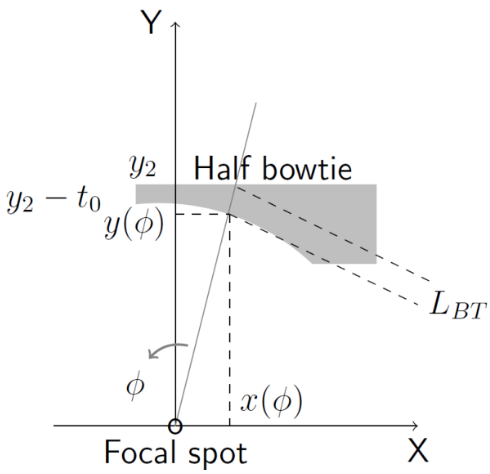

Figure 5. Coordinates of the bowtie filter's curved surface in which the y-axis coincides with the central beam axis. Y1 is the position of the filter's distal surface, t0 is the bowtie thickness through the central beam axis. ϕ is the angle between any ray through the filter and the central beam axis. The ray is transmitted through a path length LBT(ϕ) of the filter.

The resulting 2D sketch was extruded to generate a 3D model with dimensions 100.00 mm × 40.82 mm × 90.00 mm. The model was then saved as a Standard Tessellation Language (STL) file, allowing its use in defining the bowtie filter geometry for Monte Carlo simulations.

Measurement of Percentage Depth Dose (PDD) and off-axis profiles

The x-ray tube output at 100, 125, and 140 kV was characterised by measuring depth dose and off-axis dose profiles in water. A PTW BeamScan water phantom was positioned at a 100 cm Source-to-Surface Distance (SSD), with the collimator blades set to 10 cm × 28 cm, the default aperture for kV-CBCT acquisitions.

For depth dose measurements, a PTW Advanced Markus ionisation chamber (PTW Freiburg, Germany) was used, while crossline dose profiles were measured with a PTW Semiflex ionisation chamber. In both cases, a second PTW Semiflex chamber served as the reference detector. Inline profile measurements were omitted, as the bowtie filter was assumed to have a flat profile in that direction.

Percentage Depth Dose (PDD) profiles were recorded from 20 cm depth to the surface, while off-axis dose profiles were measured at 2 cm depth. The results were compared with those generated by the GATE Monte Carlo simulations for verification.

Monte Carlo Simulation of the X-ray Source

Collimator Box and Source Capsule

The collimator box is a daughter volume of the gantry, measures 40 cm × 30 cm × 34 cm and houses the x-ray source, collimator blades, and half-bowtie filter. Its centre is positioned 85 cm from the gantry origin along the y-axis.

The source capsule, which contains the x-ray source, is a cylindrical volume (10 mm length, 5 mm radius) filled with air. It is centered 15 cm from the collimator box center along the y-axis and 100 cm from the gantry isocentre. The capsule is rotated 270° about its x-axis, ensuring that the source emits radiation along the y-axis toward the isocentre.

X-ray Source

The nominal kV-CBCT energies (100, 125, and 140 kV) were characterised using discrete spectra derived from output measurements and SPEKTR 3.0, as previously described. Each spectrum was stored as a UserSpectrum text file and defined as a General Particle Source (gps) in the simulation. The source emits gamma particles with energy distributions based on the user spectrum file.

The x-ray source was modeled as a square plane (0.8 mm × 0.8 mm), centered within the source capsule. The emitted photon fluence was isotropic, with an emission angle of 14° and an azimuthal angle of 360°.

Upper Collimator Blades

The upper collimator blades define the inferior-superior field size and consist of two independently moving rectangular blades (30 cm × 10 cm × 3 mm) made of lead. They are positioned 5.91 cm from the center of the collimator box in a plane perpendicular to the y-axis.

The +Z and -Z blades are initially displaced +5 cm and -5 cm, respectively, so that their edges abut along the x-axis at zero-collimator opening. Further displacement of either blade opens the field in the z-direction.

Lower Collimator Blades

The lower collimator blades control the left-right (lateral) field size and are independently adjustable along the x-axis. They are rectangular, measuring 10 cm × 30 cm.

- The +X blade is 1 mm thick and made of stainless steel (SS304).
- The -X blade is 3 mm thick and made of uranium.

Both blades are positioned 4.31 cm from the center of the collimator box in a plane perpendicular to the y-axis. At zero-collimator opening, the edges meet along the z-axis. Further displacement defines an open field in the x-direction.

Half-Bowtie Filter

The half-bowtie filter (HBT) was modeled using FreeCAD and saved in STL format. Its bounding box dimensions are 100.00 mm × 40.82 mm × 90.00 mm. The filter is translated -0.2 mm in x and 45 mm in z, with a source-to-proximal surface distance of 14.79 cm.

In the vertical orientation, the curved side faces downstream of the source, while in the lateral orientation, it is aligned along the +X axis. The filter is made of aluminium. **Figure 6** shows a DAWN (Drawer for Academic Writings) rendering of the beamline geometry, highlighting components interacting with primary radiation.

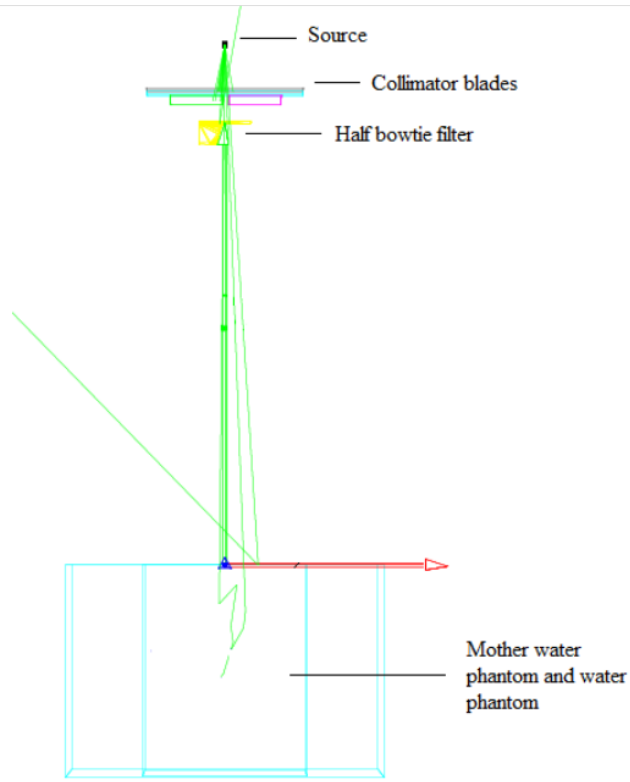

Figure 6. A DAWN visualisation of the source capsule, collimator blades, half bowtie filter, and water phantom in the simulation geometry.

Water Phantom Output

For each kV setting, Percentage Depth Dose (PDD) and crossline dose profiles were extracted from the water phantom using dose actors.

- The depth dose actor had a voxel resolution of 1 × 101 × 1 (x, y, z), where y corresponded to the number of ionisation chamber sampling points in the depth direction.
- The dose profile actor, positioned at 2 cm depth, had voxel resolutions of:
  - 100 kV: 164 × 1 × 1
  - 125 kV: 156 × 1 × 1
  - 140 kV: 162 × 1 × 1

These resolutions corresponded to the number of ionisation chamber sampling points in the x-axis. Profile scans were taken from -160 mm to 160 mm.

Physics Processes

The simulation included standard electromagnetic physics models relevant to diagnostic photon energies, implemented using the emstandard_opt3 process. This model accounts for:

- Photoelectric Effect
- Compton Scattering
- Electron Ionisation
- Bremsstrahlung

Simulation Environment

Simulations were executed using GATE v9.3, deployed in a Docker container on Microsoft Azure. The virtual machine ran Ubuntu Linux 22.04.03 ("Jammy Jellyfish"), with:

- 2 virtual CPUs (64-bit architecture)
- 8 GB RAM

Each simulation processed 8×10^8 particle histories, with a typical runtime of 6 hours per simulation.

Table 1. Measured first and second half-value layers, output and ripple for the 100, 125, and 140kV beams on the Halcyon kV-CBCT beamline.

Tube potential (kVp)	HVL1 (mm Al)	HVL2 (mm Al)	Output (mGy/mAs)	Ripple (%)
100	4.1	11.1	0.044	3.6
125	8.6	19.8	0.069	5.0
140	9.1	20.8	0.033	6.0

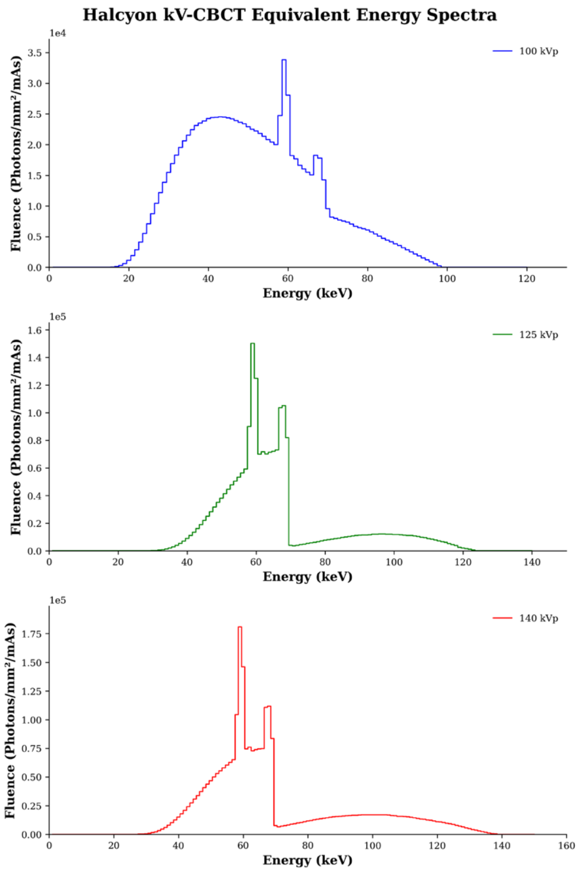

Figure 7. SPEKTR computed energy spectra of the 100kV (top), 125kV (middle), and 140kV (bottom) beams used on Halcyon for kV CBCT imaging.

Results and discussion

Beam Quality and Energy Spectra

Table 1 presents the measured HVL values, tube output (normalised to mAs), and calculated % ripple for the three x-ray beam qualities investigated. **Figure 7** illustrates the energy spectra for 100, 125, and 140 kV beams at the isocentre (100 cm from the source) as computed by SPEKTR. The corresponding mean energies were:

- 56.78 keV for 100 kV,
- 69.75 keV for 125 kV, and
- 69.95 keV for 140 kV.

Interestingly, the mean energy difference between the 125 kV and 140 kV beams was minimal (0.20 keV), reflecting in the HVL1 and HVL2 values, which differed by 0.5 mm Al and 1.0 mm Al, respectively. As expected, increasing kVp resulted in a shift toward higher mean energies and peak intensities.

Bowtie Filter Profile and 3D Model

Figure 8 displays the bowtie filter profile, derived from ion chamber measurements and analytical calculations. The corresponding 3D model, created in FreeCAD, is rendered in **Figure 9** with dimensions 100 mm × 90 mm × 40 mm.

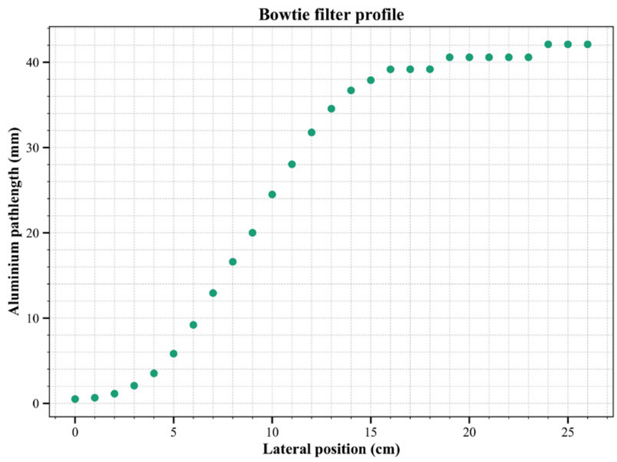

Figure 8. Profile of the curved portion of the half bowtie filter on the Halcyon kV-CBCT system. The filter material was assumed to be pure Aluminium.

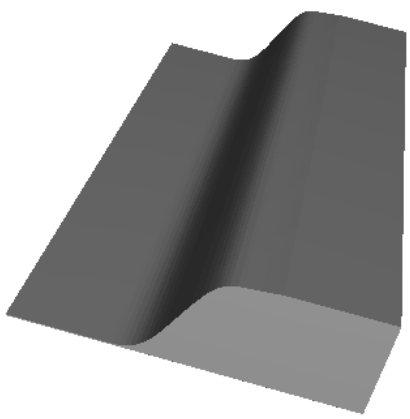

Figure 9. A 3D rendering of the half bowtie filter in the Halcyon kV-CBCT beamline created in FreeCAD.

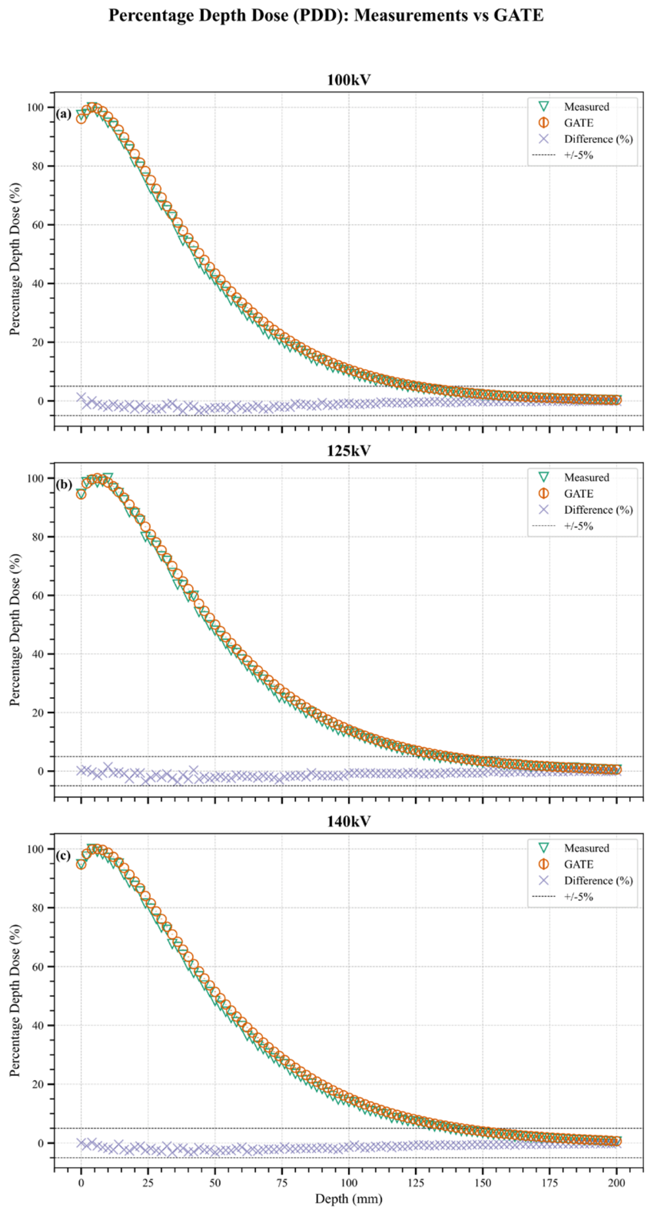

Figure 10. Comparison of measured and GATE Monte Carlo computed percentage depth dose profiles of the 100kV (a), 125kV (b), and 140kV (c) x-ray beams.

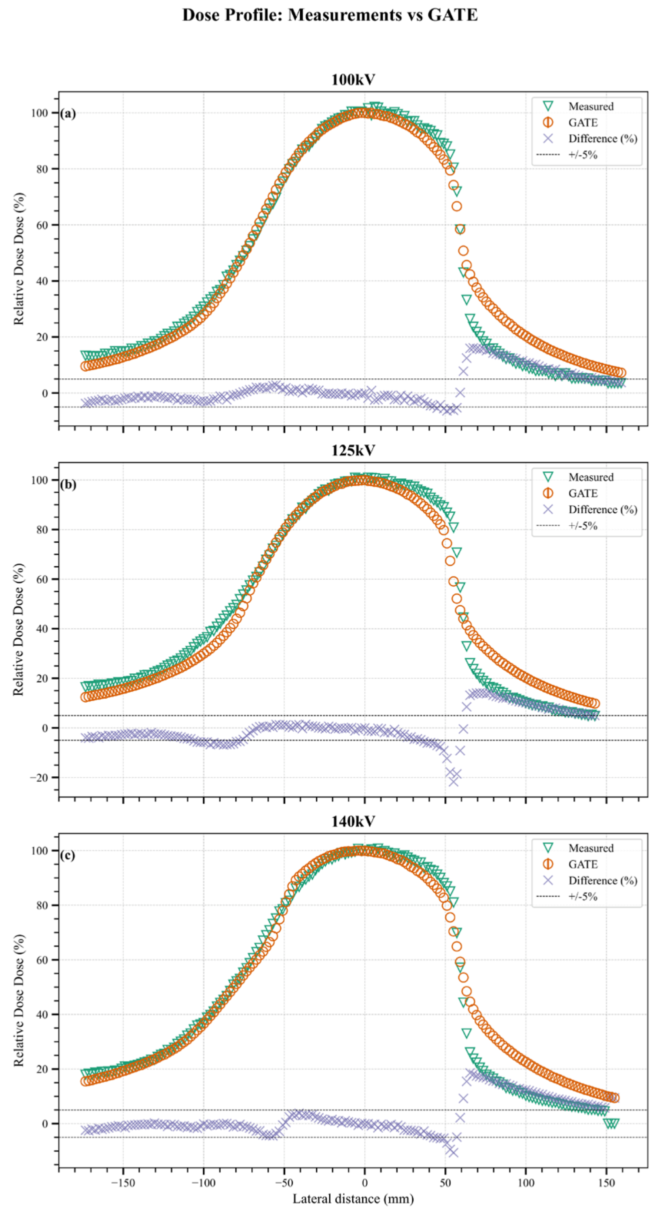

Figure 11. Comparison of measured and GATE computed crossline dose profiles of the 100kV (a), 125kV (b), and 140kV (c) x-ray beams.

Depth Dose Validation

Figure 10 compares measured and GATE-simulated PDD profiles, demonstrating maximum differences of:

- 3.47% for 100 kV,
- 3.65% for 125 kV, and
- 3.27% for 140 kV.

These differences were most pronounced in the dose fall-off region, where GATE slightly overestimated the percentage dose. Depth-dose data were normalised to the dose maximum at each energy, and the strong agreement suggests that the spectral modeling was accurate and reliable.

Crossline Dose Profile and Filtration Model Validation

The crossline dose profiles, shown in **Figure 11**, were used to evaluate the accuracy of the filtration model, particularly the bowtie filter representation. Across all beam energies, the maximum discrepancies within the central beam and beneath the curved section of the bowtie filter were:

- 3.73% for 100 kV,
- 6.84% for 125 kV, and
- 4.44% for 140 kV.

Most data points were within 3%, confirming that the bowtie filter model accurately represented beam attenuation and shaping. However, differences became more pronounced in the penumbra region on the flat side of the bowtie filter:

- 15.95% for 100 kV,
- 21.72% for 125 kV, and
- 18.57% for 140 kV.

These large deviations in the high-gradient penumbra regions are attributed to mismatches in spatial resolution between the measurement detectors and the GATE scoring geometry.

Detector Limitations and Spatial Resolution Constraints

A Farmer-type ionisation chamber (0.65 cc sensitive volume) was used for in-air bowtie filter profiling, owing to its high signal-to-noise ratio (SNR). While a smaller-volume chamber would have improved spatial resolution, it would have required higher tube loading, increasing the risk of x-ray tube overloading. To mitigate this, the smaller chamber was instead used for in-water dose distribution profiling.

The voxel dimensions for scoring the off-axis and depth dose profiles were carefully matched to the ionisation chamber sampling points in the water phantom. While depth dose resolutions were well-aligned, scoring voxels were larger than the chamber volume for profile measurements, leading to partial volume effects. A limitation of GATE v9.3 was that scoring volume resolution could only be defined by voxel number or voxel size, but not both simultaneously. If voxel sizes were matched to the detector volume, it would cause scaling mismatches due to discrepancies in the number of sampling points.

For reference, Li et al.[6] reported maximum differences of 25.9% between Monte Carlo-simulated dose distributions and in vitro measurements, despite using a more rigorous bowtie filter profiling approach which involved energy-specific bowtie profiles.

Conclusion

This study presents equivalent source and filtration models of the kV-CBCT system onboard the Halcyon 2.0 linear accelerator for 100, 125, and 140 kV beam energies commonly used for adult imaging . Due to the lack of publicly available manufacturer-supplied geometry and source data, these models were developed using semi-empirical methods.

Validation against measured PDDs in a water phantom showed differences within 3.7%, while crossline dose profiles within the central beam differed by up to 6.9%. However, in high-gradient penumbral regions, discrepancies reached 22%, primarily due to the use of detectors and dose scorers with high signal-to-noise ratios (SNR) but limited spatial resolution.

Despite these localised differences, the overall impact on patient dosimetry accuracy is expected to be minimal, supporting the feasibility of using these equivalent models for Monte Carlo-based kV imaging dose calculations on the Halcyon.

References

1. Stanley DN, Harms J, Pogue JA, et al. A roadmap for implementation of kV-CBCT online adaptive radiation therapy and initial first year experiences. J Applied Clin Med Phys. 2023;24(7). doi:10.1002/acm2.13961
2. Ding GX, Alaei P, Curran B, et al. Image guidance doses delivered during radiotherapy: Quantification, management, and reduction: Report of the <scp>AAPM</scp> Therapy Physics Committee Task Group 180. Medical Physics. 2018;45(5). doi:10.1002/mp.12824
3. Rogers DWO. Fifty years of Monte Carlo simulations for medical physics. Phys Med Biol. 2006;51(13):R287-R301. doi:10.1088/0031-9155/51/13/r17
4. Turner AC, Zhang D, Kim HJ, et al. A method to generate equivalent energy spectra and filtration models based on measurement for multidetector CT Monte Carlo dosimetry simulations. Medical Physics. 2009;36(6Part1):2154-2164. doi:10.1118/1.3117683
5. McMillan K, McNitt-Gray M, Ruan D. Development and validation of a measurement-based source model for kilovoltage cone-beam CT Monte Carlo dosimetry simulations. Medical Physics. 2013;40(11). doi:10.1118/1.4823795
6. Li X, Shi JQ, Zhang D, et al. A new technique to characterize CT scanner bow-tie filter attenuation and applications in human cadaver dosimetry simulations. Medical Physics. 2015;42(11):6274-6282. doi:10.1118/1.4932364
7. Altergot A, Schürmann M, Jungert T, et al. Imaging doses for different CBCT protocols on the Halcyon 3.0 linear accelerator – TLD measurements in an anthropomorphic phantom. Zeitschrift für Medizinische Physik. 2024;34(4):580-595. doi:10.1016/j.zemedi.2023.03.002
8. Gazdag-Hegyesi S, Gáldi Á, Major T, Pesznyák C. Dose indices of kilovoltage cone beam computed tomography for various image guided radiotherapy protocols. Radiation Protection Dosimetry. 2023;199(8-9):983-988. doi:10.1093/rpd/ncad101
9. Punnoose J, Xu J, Sisniega A, Zbijewski W, Siewerdsen JH. Technical Note: spektr 3.0-A computational tool for x-ray spectrum modeling and analysis. Med Phys. 2016;43(8Part1):4711-4717. doi:10.1118/1.4955438
10. Hernandez AM, Boone JM. Tungsten anode spectral model using interpolating cubic splines: Unfiltered x-ray spectra from 20 kV to 640 kV. Medical Physics. 2014;41(4). doi:10.1118/1.4866216
11. Zhang G, Marshall N, Jacobs R, Liu Q, Bosmans H. Bowtie filtration for dedicated cone beam CT of the head and neck: a simulation study. BJR. 2013;86(1028):20130002. doi:10.1259/bjr.20130002
12. Mayer JR, 20 January 2024. [Online]. Available: https://www.freecadweb.org.