

ORIGINAL RESEARCH ARTICLE

Monte Carlo characterization of a cabinet X-ray irradiator using PHITS 3.35: Dosimetry, half-value layer assessment, and implications for pediatric facility shielding

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Abstract

Accurate characterization of X-ray beams underpins radiation protection, dosimetry, and shielding design. This study presents a Monte Carlo simulation framework using the Particle and Heavy Ion Transport Code System (PHITS) 3.35 to characterize the photon energy spectra of a cabinet X-ray irradiator (Hitachi MBR1618RBE) at 80, 100, and 120 kVp, spanning the voltage range used for pediatric chest and abdominal radiography. A tungsten anode tube model (20° anode, 0.8 mm Be window, 0.5 mm Al + 0.1 mm Cu filter) was implemented, with spectral fluence scored via the TTrack tally at $z = 7$ cm (2×10^7 histories/run). Simulated mean photon energies were 57.9, 62.0, and 67.6 keV. First half-value layers (HVL_1), derived from National Institute of Standards and Technology Photon Cross Sections data, were 3.08, 3.25, and 3.43 mm Al, with homogeneity coefficients of 0.970–0.989 confirming polychromatic beam quality. Depth-dose, off-axis, entrance surface air kerma, effective dose, and lead-shielding quantities were derived by analytical spectral convolution and published coefficients rather than full Monte Carlo transport, yielding bounding, illustrative estimates. Pediatric effective dose (5-year-old phantom, source-to-surface distance = 25 cm, National Radiological Protection Board W14 factors) was 0.35–5.19 mSv per examination; these are not directly comparable to clinical diagnostic reference levels. Required lead thicknesses at a 1.5 m wall under conservative occupancy assumptions were 8.3–15.3 mm Pb (public) and 5.6–11.3 mm Pb (occupational). A PHITS implementation note distinguishing vacuum (code 0) from outer void (code 1) is documented for practitioners. No independent experimental validation was performed. These results characterize the simulated source term and illustrate a PHITS-based workflow, not validated clinical reference values.

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Citation: Saragih RB. Monte Carlo characterization of a cabinet X-ray irradiator using PHITS 3.35: Dosimetry, half-value layer assessment, and implications for pediatric facility shielding. *Adv Radiother Nucl Med.* 2026;4(3):026220019.
doi: 10.36922/ARNM026220019

Received: May 30, 2026

Revised: August 31, 2026

Accepted: September 4, 2026

Published online: September 29, 2026

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Keywords: Monte Carlo; Particle and Heavy Ion Transport code System; X-ray energy; Half-value layer; Cabinet irradiator; Pediatric dosimetry; Lead shielding

1. Introduction

X-ray irradiators are preferred in many radiobiology, pharmacology, and cancer research settings due to several practical advantages. Small X-ray radiation platforms have advanced to enable precise image-guided irradiation protocols that facilitate rigorous radiobiological investigations, including studies that compare X-ray effects

to other radiation modalities such as protons.¹ Despite their widespread use and convenience, dose accuracy and standardization in X-ray irradiators remain persistent challenges. A survey of dosimetry methods across multiple laboratories revealed significant variability in dose delivery, with only one out of five facilities using X-ray irradiators achieving within 5% of the target dose, while others ranged from 12% to 42% deviation.² This variability underscores the need for standardized dosimetry practices and stringent quality assurance protocols to ensure reproducibility and reliability in radiobiological experiments, an essential step to improve accuracy and support the translational potential of preclinical research findings.³

Diagnostic X-ray examinations represent the largest artificial source of population radiation exposure globally, contributing approximately 20% of per capita effective dose from all sources.⁴ In pediatric patients, this concern is amplified because children exhibit 23 times greater radiosensitivity than adults, attributable to higher cellular division rates, larger organ-to-body size ratios, and longer remaining life expectancy over which stochastic effects may manifest.^{5,6} The as low as reasonably achievable principle therefore demands precise characterization of the radiation beams used in clinical practice, both to protect patients and to enable accurate shielding design for facility personnel and the public.

X-ray spectra are shaped by two physical mechanisms. Bremsstrahlung produces a continuous energy distribution from near zero to the peak tube voltage in kVp, while characteristic emission from tungsten K-shell transitions, predominantly Ka1 at 59.3 keV and Kb1 at 67.2 keV, appears as discrete peaks superimposed on the continuum.⁷ Both components are modified by tube filtration, including the beryllium window and external Al and Cu filters, which selectively attenuate low-energy photons and harden the beam. The resulting spectral shape primarily determines beam interaction with biological tissue, governing absorbed dose distribution, penetration depth, and image contrast.

Monte Carlo simulation is the reference method for X-ray beam characterization, accurately modeling photon–electron interaction physics without requiring physical measurement apparatus.⁸ Among available Monte Carlo platforms, including Electron Gamma Shower of the National Research Council of Canada (EGSnrc), Monte Carlo N-Particle, Geant4, and Particle and Heavy Ion Transport code System (PHITS), developed by the Japan Atomic Energy Agency, provides comprehensive kilovoltage photon transport via the Electron Gamma Shower Version 5 (EGS5) engine and is increasingly adopted in Asian medical physics centers.^{9,10} Notwithstanding detailed

EGSnrc-based characterization studies of cabinet X-ray irradiators,^{11,12} a PHITS-based framework covering the diagnostic voltage range of 80, 120 kVp with an integrated, downstream dosimetry and shielding estimation pipeline has not been reported. A previous study independently characterizes the same Hitachi MBR1618RBE irradiator using EGSnrc¹¹; the present study differs in using PHITS/EGS5 for the spectral simulation and in adding a code-agnostic analytical pipeline (half-value layer [HVL], depth dose/offaxis convolution, effective dose estimation, and shielding design) built on the simulated spectrum, rather than in presenting new Monte Carlo dose calculations in a phantom or room. This gap is significant for PHITS users because the code is widely used for regulatory and research purposes across the region, including for cabinet irradiator and small animal irradiator characterization.

This study therefore aims to: (i) Simulate diagnostic X-ray spectra at 80, 100, and 120 kVp with PHITS 3.35 and EGS5 physics; (ii) compute halfvalue layers from simulated spectra using National Institute of Standards and Technology (NIST) Photon Cross Sections (XCOM) attenuation data; (iii) derive depth dose and offaxis dose profiles in a water phantom by analytical spectral convolution; (iv) estimate effective dose for a 5 year old patient via National Radiological Protection Board (NRPB) W14 normalized dose factors; and (v) determine Pb shielding thicknesses following National Council on Radiation Protection and Measurements (NCRP) 49 and International Atomic Energy Agency (IAEA) Basic Safety Standards. The results are intended to provide an open, reproducible PHITS-based workflow for spectral simulation, beam quality assessment, and illustrative dose/shielding estimation for this cabinet irradiator, with potential relevance to radiation protection practice pending independent validation.

2. Materials and methods

2.1. X-ray tube model and geometry

The X-ray tube model follows the specifications of the Hitachi MBR1618RBE irradiation system (Hitachi Power Solutions Co., Ltd., Japan), a widely used cabinet X-ray unit in biomedical research.¹¹ The geometrical model was implemented in PHITS 3.35 using planar and cylindrical surface primitives with the focal spot at $z = 0$ cm. A monoenergetic parallel circular electron beam (stype = 1, radius = 0.5 mm, corresponding to a 1 mm focal spot diameter) was directed along the positive z -axis onto a 0.3 mm tungsten slab (density 19.3 g/cm³, $Z = 74$), representing the anode target. The 20° anode angle was accounted for by the effective target thickness along the beam direction; this flat-slab representation does not model angular photon

self-absorption within the target, the heel effect, or takeoff-angle-dependent spectral variation, and is used here as a simplified, computationally efficient approximation for this fixed takeoff geometry (Section 3.6). Key component parameters are summarized in Table 1. Note on scope: the geometry and irradiation parameters modeled (1 mm focal spot, source-to-surface distance [SSD] = 25 cm, cabinet enclosure) correspond specifically to the Hitachi MBR1618RBE preclinical/cabinet X-ray irradiator, not to a clinical diagnostic radiography room; the pediatric dose and shielding estimates in Sections 2.5, 2.6, 3.4, and 3.5 are downstream illustrative applications of the simulated spectra to this specific source and should not be read as characterizing a clinical diagnostic imaging system.

2.2. Particle and Heavy Ion Transport Code System simulation setup

All simulations were performed with PHITS version 3.35 using the EGS5 physics module (negs = 1) for coupled electron-photon transport. Energy cutoffs of 1 keV were applied to electrons, positrons, and photons. Physical processes modeled included bremsstrahlung emission (Bethe–Heitler cross sections), photoelectric absorption, Compton scattering (Klein–Nishina), Rayleigh scattering, and characteristic X-ray emission following K-shell ionization; photon interaction cross sections were obtained from the XCOM database.¹³ A total of 2×10^7 primary electron histories per run (20 batches $\times$ 10^6 histories each) were simulated to achieve statistical uncertainties below 1% in the central beam region. These are the per-energy-bin statistical uncertainties (one relative standard deviation) reported by the TTrack tally; derived quantities (HVL, mean energy, dose) inherit comparable uncertainty through error propagation and were not separately resampled or bootstrapped in this study. A practical implementation detail relevant to PHITS users, rather than a novel physics finding, is the distinction between PHITS material code 0, which represents vacuum and allows free particle transport, and code 1, which denotes the outer void and immediately terminates particles. Incorrect

assignment of 1 to interior vacuum cells yields null dose output with no error message, a geometry pitfall distinct from EGSnrc conventions that must be carefully observed when adapting models between codes. Throughout this study, photon fluence $\Phi(E)$ (photons $\text{cm}^2 \text{MeV}^{-1}$) from the simulation is used directly for the HVL calculation (Section 2.3); the same fluence, weighted by energy and the water mass energy absorption coefficient, is used for the depth-dose and off-axis calculations (Section 2.4); and published tube output and entrance surface air kerma (ESAK) tables, not the simulated fluence, are used for the effective dose and shielding calculations (Sections 2.5 and 2.6). This distinction is made explicit because fluence-based and kerma-based quantities are not interchangeable.

Photon spectral fluence was scored using the TTrack tally in region-based mesh mode (mesh = reg) within a cylindrical air volume of radius 2.5 cm and depth 0.10 cm centered at $z = 7$ cm. Energy binning employed 200 linear bins from 1 keV to the maximum photon energy equal to the tube voltage in keV. Output fluence was normalized per source electron in units of photons $\text{cm}^2 \text{MeV}^{-1}$, with the scoring volume of 1.964 cm^3 explicitly declared for correct per cm^2 normalization.

2.3. Half-value layer calculation

Half-value layers were calculated analytically from the simulated spectra using the polychromatic transmission integral. The transmitted intensity fraction through an aluminum absorber of thickness x (cm) was computed as in Equation 1:

$$I(x)/I_0 = \int \Phi(E) \cdot \exp[-\mu_{\text{Al}}(E) \cdot x] dE / \int \Phi(E) dE \quad (1)$$

where $\Phi(E)$ is the simulated photon fluence spectrum and $\mu_{\text{Al}}(E)$ is the linear attenuation coefficient of aluminum, obtained as the product of the mass attenuation coefficient from NIST XCOM and the aluminum density of 2.699 g/cm^3 .¹³ The first half-value layer (HVL_1) was defined as the thickness yielding $I(x)/I_0 = 0.5$, and the second half-value layer (HVL_2) as the thickness yielding $I(x)/I_0 = 0.25$. Both

Table 1. Geometry and material parameters of the X-ray tube model

Component	Material	z range (cm)	Thickness (mm)	Density (g/cm ³)
W anode target	Tungsten (Z = 74)	0.00 0.03	3.0	19.300
Be window (inherent)	Beryllium (Z = 4)	1.92 2.00	0.8	1.848
Al filter (external)	Aluminium (Z = 13)	6.89 6.94	0.5	2.699
Cu filter (external)	Copper (Z = 29)	6.94 6.95	0.1	8.960
Scoring plane (air)	Dry air (STP)	6.95 7.05	10 (volume)	1.205×10^3

were determined using the Brent root-finding algorithm. The homogeneity coefficient was computed as in Equation 2:

$$HE = \frac{HVL_1}{HVL_2 - HVL_1} \quad (2)$$

with values below unity confirming polychromatic beam character. HVL_2 is defined here, following the International Electrotechnical Commission convention, as the total absorber thickness required to reduce transmission to 25% (i.e., inclusive of HVL_1), not as the second incremental thickness added after HVL_1 ; this is stated explicitly to avoid ambiguity with the alternative “second HVL” convention used in some shielding texts.

2.4. Depth dose and off-axis dose profile

Absorbed dose as a function of depth d in a water phantom was estimated by kerma-weighted spectral convolution: $D(d)$ is proportional to the integral (Equation 3):

$$D(d) \propto \int \Phi(E) \cdot E \cdot (\mu_{en}/\rho)_{water}(E) \cdot \exp[-\mu_{water}(E) \cdot d] dE \quad (3)$$

where $(\mu_{en}/\rho)_{water}$ is the mass energy absorption coefficient from NIST XCOM.¹³ The lateral off-axis dose profile at the phantom surface ($SSD = 250$ mm) was modeled using a flat-top geometric field projected from the collimator aperture of radius 2.5 cm at $z = 6.95$ cm, yielding a field radius at the patient surface of $r = 2.5 \times 25.0/6.95 = 9.0$ cm, combined with a Gaussian penumbra of $\sigma = 0.8$ cm to account for focal spot size and small-angle scatter contributions. These depth-dose and off-axis calculations are analytical spectral convolutions using the simulated PHITS spectra as the fluence input; they are not full three-dimensional (3D) Monte Carlo transport calculations in a phantom or patient, and heterogeneity, scatter buildup, and true divergent-beam geometry are not explicitly modeled.

2.5. Pediatric effective dose estimation

Patient effective dose was estimated for a 5-year-old child in anterior–posterior (AP) geometry using the normalized dose factor (NDF) method (Equation 4):

$$E = E_{SAK} \times NDF \quad (4)$$

where E_{SAK} is the entrance surface air kerma at the patient surface, and NDF is the normalized dose factor tabulated for the examination type, tube voltage, and patient age.¹⁴ E_{SAK} was derived from published tube output values at a reference distance of 100 cm (Institute of Physics and Engineering in Medicine [IPEM] Report 32, Part VII), scaled to $SSD = 25$ cm by the inverse square law. NDF coefficients for AP geometry and the 5-year-old

pediatric phantom were taken from NRPB W14 Table 5¹⁴: 0.134, 0.095, and 0.082 mSv/mGy at 80, 100, and 120 kVp, respectively. Clinical examination techniques were 1.0 mAs for chest posterior–anterior (PA) at 80 kVp, 5.0 mAs for abdomen AP at 100 kVp, and 8.0 mAs for abdomen AP at 120 kVp. Results were cross-validated against International Commission on Radiological Protection Publication 116 fluence-to-effective-dose conversion coefficients for AP geometry.¹⁵ For consistency with the AP projection NDF coefficients available in NRPB W14 Table 5, the chest examination is also modeled here using the AP projection dose factor as an approximation; PA-specific chest NDFs, typically 10%–20% lower than AP values for the same entrance kerma, were not applied. This simplification should be corrected using projection-specific coefficients (e.g., NRPB W14 Table 4 for PA chest) before the chest dose estimate is used for any clinical comparison.

2.6. Shielding design

Required lead shielding thickness was determined using the multi-HVL method (NCRP 49¹⁶; BIR 2000¹⁷). The weekly unshielded air kerma at the design wall was calculated as in Equation 5:

$$K_{wall} = \frac{\text{Output (100 cm)} \times \text{mAs} \times N_{views/wk}}{(100 \text{ cm} / d_{wall})^2} \quad (5)$$

where $d_{wall} = 150$ cm, the workload was 50 patients per week with two views each (100 exposures per week), and tube output values were from IPEM Report 32. For polychromatic X-ray beams, the required Pb thickness uses HVL_1 for the first decade of attenuation and HVL_2 for subsequent decades to account for beam hardening through the shield. Pb HVL values were taken from NCRP 49. Design goals followed IAEA Basic Safety Standards¹⁸: $P = 1.0$ mGy/week for occupational areas (corresponding to 50 mSv/year) and $P = 0.02$ mGy/week for public uncontrolled areas (corresponding to 1 mSv/year). NCRP 49 was used here because it tabulates Pb half-value layers directly as a function of kVp and filtration for the multi-HVL shielding method used in this modeling exercise; contemporary diagnostic room shielding design should instead follow NCRP 151¹⁹ or an equivalent current national standard, and the results in Section 3.5 should be treated as illustrative bounding estimates rather than facility design values.

3. Results and discussion

3.1. Simulated energy spectra

The simulated photon energy spectra at the scoring plane ($z = 7$ cm) are shown in Figure 1 for all three tube voltages. Each spectrum presents the expected bremsstrahlung

continuum rising from near zero energy to a peak in the 4,070 keV range, followed by an abrupt cutoff at the maximum photon energy equal to the tube voltage in keV. Distinct tungsten characteristic emission lines are clearly resolved: the Ka doublet (Ka2 at 57.98 keV and Ka1 at 59.32 keV) and the Kb group (Kb1 at 67.24 keV and Kb2 at 69.10 keV) are superimposed on the bremsstrahlung background at all three voltages. The number of energy bins with nonzero fluence increased with kVp, from 44 bins at 80 kVp to 48 at 100 kVp and 67 at 120 kVp, consistent with the broader spectral range at higher voltages. Peak fluence increased monotonically with kVp, reflecting the greater bremsstrahlung yield at higher electron energies. Fluence-weighted mean photon energies were 57.9, 62.0, and 67.6 keV at 80, 100, and 120 kVp, respectively, and are summarized with other beam quality parameters in [Table 2](#).

3.2. Half-value layer and beam quality

[Table 2](#) presents the beam quality parameters derived from the simulated spectra. HVL_L increased from 3.08 mm Al at 80 kVp to 3.43 mm Al at 120 kVp, reflecting the progressively harder beam quality as the spectral mean shifts to higher energies at greater tube voltages; these values fall within the ~2.55 mm Al range reported for similarly filtered 80,120 kVp diagnostic beams in the

literature. HVL_L values were 6.26, 6.57, and 6.90 mm Al at 80, 100, and 120 kVp, respectively. The homogeneity coefficient was systematically below unity at all voltages (0.970, 0.978, and 0.989), confirming polychromatic beam character, and approached unity at higher kVp as high-energy photons increasingly dominated the spectrum. This progression has a practical implication for this simulated beam: beam hardening corrections are more important for 80 kVp chest-equivalent protocols than for 120 kVp abdomen-equivalent settings, where the beam is comparatively more energetically homogeneous. The transmission curves, HVL comparison, spectral hardening after passage through HVL_L , and the homogeneity coefficient trend are illustrated in [Figure 2](#).

3.3. Depth dose and off-axis profiles

As noted in Section 2.4, the depth-dose and off-axis results presented here are analytical spectral convolutions derived from the simulated PHITS spectra, not full 3D Monte Carlo dose calculations in a phantom. The percentage depth dose profiles along the central axis and the lateral off-axis dose distribution at the phantom surface (SSD = 250 mm) are presented in [Figure 3](#). All three beams exhibited the characteristic kilovoltage dose pattern of maximum dose at the surface followed by exponential attenuation with increasing depth, in contrast to the buildup region

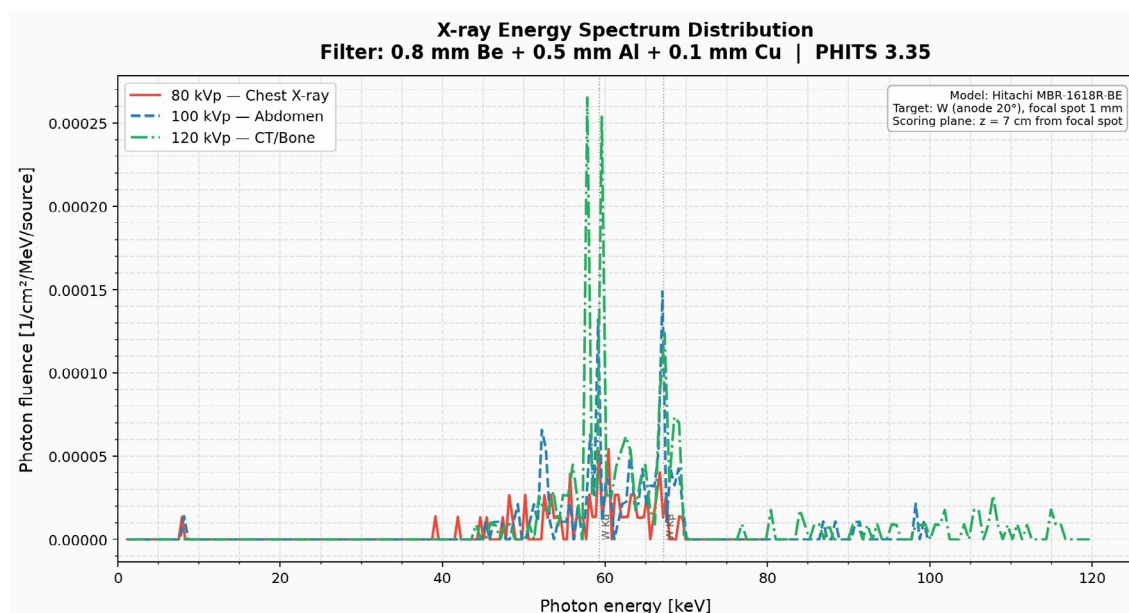


Figure 1. Simulated X-ray energy spectra at the scoring plane ($z = 7$ cm from focal spot) for 80 kVp (chest PA), 100 kVp (abdomen AP), and 120 kVp (higher kVp abdomen/pelvis). Filter combination: 0.8 mm Be + 0.5 mm Al + 0.1 mm Cu. Tungsten Ka and Kb characteristic emission lines are annotated. Simulated with PHITS 3.35/EGS5.

Abbreviations: AP: Anterior–posterior; CT: Computed tomography; PA: Posterior–anterior.

Table 2. Spectral and beam quality parameters derived from Particle and Heavy Ion Transport Code System 3.35 simulations for 80, 100, and 120 kVp

Parameter	80 kVp	100 kVp	120 kVp	Unit
Mean photon energy	57.9	62.0	67.6	keV
Nonzero fluence bins	44	48	67	
HVL ₁	3.08	3.25	3.43	mm Al
HVL ₂	6.26	6.57	6.90	mm Al
Homogeneity coefficient, HE	0.9702	0.9780	0.9894	

Abbreviation: HVL: Half-value layer.

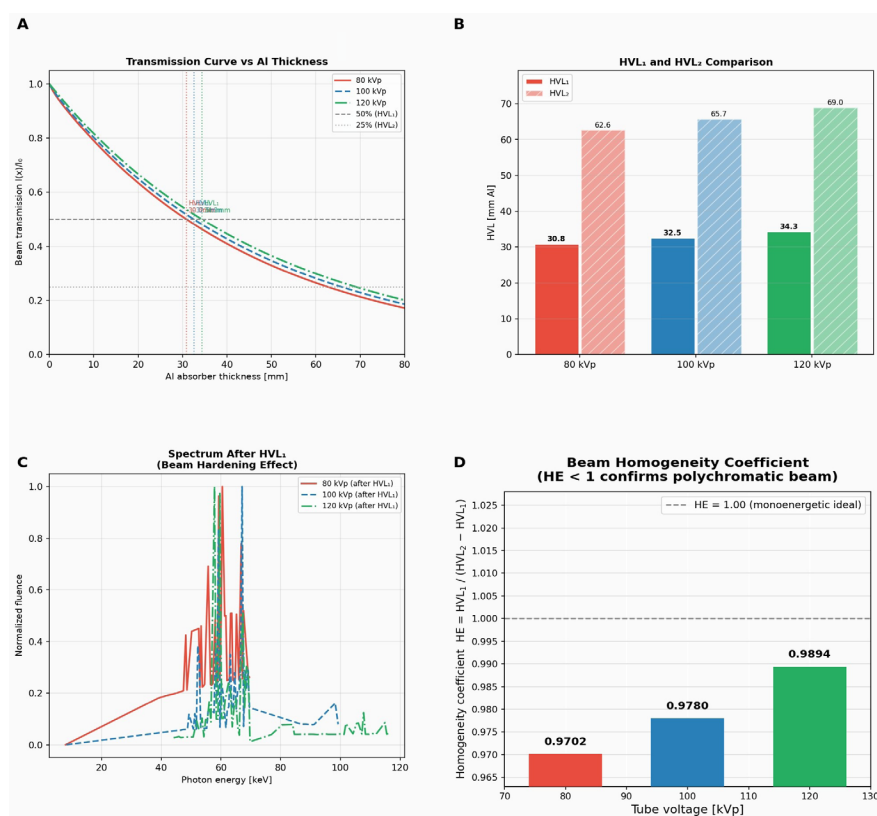


Figure 2. Half-value layer (HVL) analysis. (A) Polychromatic aluminum transmission curves with HVL₁ positions indicated. (B) HVL₁ and HVL₂ comparison across kVp. (C) Spectral shift after passage through HVL₁ demonstrating beam hardening. (D) Homogeneity coefficient (HE) as a function of tube voltage.

observed for megavoltage beams. Percentage depth dose at 5 cm depth was approximately 44%, 47%, and 51% for 80, 100, and 120 kVp, respectively; the increase at higher voltage reflects the lower water attenuation coefficient of more energetic photons. This depth-dose difference is

relevant when comparing radiation penetration for the abdomen-equivalent versus chest-equivalent settings modeled here.

The off-axis profiles show a flat-top region extending to approximately 9.0 cm radius, corresponding to the

geometric projection of the 2.5 cm collimator aperture at a magnification factor of $SSD/z_{coll} = 25.0/6.95 = 3.60$. A sharp penumbra at the field edge characterizes all three voltages, with a measured 80%–20% penumbra width of 1.11 cm, independent of kVp, consistent with the fixed 1 mm focal spot size and constant scatter conditions. The essentially voltage-independent profile shape confirms that beam geometry, rather than spectral composition, governs lateral dose uniformity under these irradiation conditions.

3.4. Pediatric patient effective dose

Table 3 presents the ESAK and effective dose estimates for each examination type in a 5-year-old child. ESAK per view ranged from 1.31 mGy for chest PA at 80 kVp to 31.62 mGy for abdomen AP at 120 kVp, with the large difference primarily reflecting the higher mAs required for adequate abdominal penetration. Effective dose per examination (two views) was 0.35, 2.37, and 5.19 mSv at 80, 100, and 120 kVp, respectively. Table 3 also lists indicative pediatric diagnostic reference level (DRL) values from EC Report

EUR 16262 for context.¹⁷ DRLs are conventionally defined in terms of entrance dose quantities (entrance surface dose or air kerma) measured at clinical SSDs, not the effective dose quantity, short SSD geometry, and simplified anode/collimation model used in this study; a direct DRL exceedance comparison is therefore not clinically valid here. For example, the 80 kVp chest effective dose of 0.35 mSv numerically exceeds the listed 0.10 mSv chest DRL value, illustrating that the two quantities are not interchangeable; no DRL compliance conclusion should be drawn from Table 3. The steep increase from chest to abdominal protocols underscores the importance of using the minimum clinically adequate technique, particularly for repeated examinations in pediatric oncology patients. At the SSD of 25 cm used in this irradiator geometry, effective dose values lie at the upper boundary of published pediatric survey data; clinical SSDs of 100–180 cm used in diagnostic rooms would reduce ESAK by factors of 16–52, yielding effective doses well below 0.1 mSv for chest examinations.

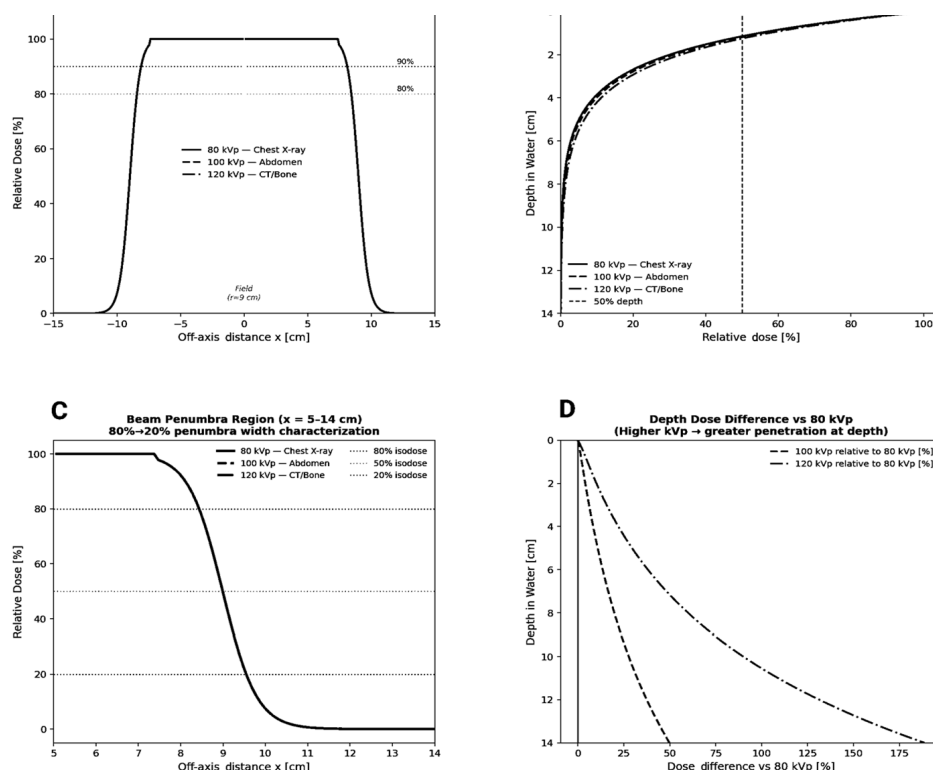


Figure 3. Dose distribution results at SSD = 250 mm in a water phantom. (A) Off-axis dose profiles at the phantom surface. (B) Percentage depth dose along the central axis. (C) Penumbra region detail showing the 80%–20% transition. (D) Relative depth dose increase of 100 and 120 kVp compared to 80 kVp. Profiles derived from analytical spectral convolution using PHITS 3.35 spectra and NIST XCOM water attenuation coefficients.

Abbreviations: CT: Computed tomography; NIST: National Institute of Standards and Technology; PHITS: Particle and Heavy Ion Transport Code System; SSD: Source-to-surface distance; XCOM: Photon Cross Sections.

Table 3. Entrance surface air kerma (ESAK) and effective dose estimates for a 5-year-old child in anterior–posterior (AP) geometry

Parameter	80 kVp chest PA	100 kVp abdomen	120 kVp abdomen	Unit
mAs per view	1.0	5.0	8.0	mAs
ESAK per view	1.31	12.48	31.62	mGy
NDF (5 yr, AP, NRPB W14)	0.134	0.095	0.082	mSv/mGy
Effective dose per exam	0.35	2.37	5.19	mSv
Indicative pediatric DRL (entrance dose-based; not directly comparable, see text)	0.10	~5	~510	mSv
DRL exceeded (not evaluable; see text)	N/A	N/A	N/A	–

Notes: NDF values from Hart *et al.*¹⁴ NRPB W14 Table 5; DRL values from EC Report EUR 16262.¹⁷

Abbreviations: AP: Anterior–posterior; DRL: Diagnostic reference level; ESAK: Entrance surface air kerma; NDF: Normalized dose factor; NRPB: National Radiological Protection Board; PA: Posterior–anterior.

3.5. Lead shielding requirements

Table 4 summarizes the shielding calculation results. Unshielded weekly air kerma at the 1.5 m wall increased from 33.2 uGy/week at 80 kVp to 1260 uGy/week at 120 kVp, a 38-fold increase over the voltage range studied. Required Pb thicknesses for public uncontrolled areas (design goal 0.02 mGy/week) ranged from 8.3 mm at 80 kVp to 15.3 mm at 120 kVp; for occupational areas (design goal 1.0 mGy/week), requirements were 5.611.3 mm. These values appear high relative to the 1.53 mm Pb conventionally installed in diagnostic rooms, but the calculation assumes conservative worst-case occupancy ($T = 1$) and use factors ($U = 1$). Applying facility-specific occupancy (T) and use (U) factors, typically well below 1 for corridors, external walls, and lightly occupied areas, would reduce the required thickness substantially; however, the magnitude of the reduction depends on the specific T and U values applicable to each barrier and was not recalculated in this study. Convergence with the 1.53 mm Pb typically installed in diagnostic rooms should therefore be verified by a qualified radiation protection expert applying facility-specific T and U factors following current guidance (e.g., NCRP 151 [22]), rather than assumed from the bounding estimate reported here. The 120 kVp condition governs shielding design when multiple voltages are used, as any wall thickness adequate for 120 kVp automatically satisfies the requirements for 80 and 100 kVp. Shielding metrics and Pb transmission curves are shown in Figure 4.

3.6. Limitations

Several limitations should be considered when interpreting

these results. First, the anode is modeled as a flat 0.3 mm tungsten slab with an effective thickness correction for the 20° takeoff angle; this simplification does not capture angular photon self-absorption, the heel effect, or takeoff-angle-dependent spectral hardening, and should be validated against a full 3D target geometry or measurement before use in applications sensitive to off-axis spectral shape. Second, no independent experimental or code-to-code validation of the simulated spectra or derived HVLs was performed; comparison with measured HVL data, manufacturer specifications, or an independent spectrum generation code such as SpekPy²⁰, or catalogue reference spectra (e.g., IPEM Report 78/SRS78³), would strengthen confidence in the results, as would comparison with dosimetric characterizations of dedicated pediatric X-ray units such as that of Smans *et al.*²¹ Third, the depth dose, off-axis, effective dose, and shielding results are derived from the simulated spectra by analytical convolution and tabulated published coefficients rather than by full Monte Carlo transport through a voxelized patient or room geometry; they should be read as source term characterization and illustrative dose/shielding estimates for this specific cabinet irradiator, not as clinical dosimetry or facility shielding design values. Fourth, effective dose was estimated using the AP projection NRPBW14 normalized dose factor for all three examinations, including the 80 kVp chest case, which is more commonly performed PA; projection-specific coefficients (and, where relevant, updated International Commission on Radiological Protection 103²² tissue weighting factors) should be applied before any clinical comparison is drawn. Fifth, the reported lead

Table 4. Lead shielding requirements at wall distance $d = 150$ cm, workload 50 patients/week $\times$ 2 views

Parameter	80 kVp	100 kVp	120 kVp	Unit
Pb HVL1 (NCRP 49)	0.24	0.27	0.31	mm Pb
Pb HVL2 (NCRP 49)	0.49	0.58	0.71	mm Pb
K _{wall} unshielded	33.2	504	1260	uGy/week
Pb required, occupational	5.55	8.44	11.26	mm Pb
Pb required, public	8.31	11.71	15.27	mm Pb

Notes: Pb half-value layer (HVL) values from National Council on Radiation Protection and Measurements (NCRP) 49¹⁶; dose limits from International Atomic Energy Agency BSS (2014).¹⁸

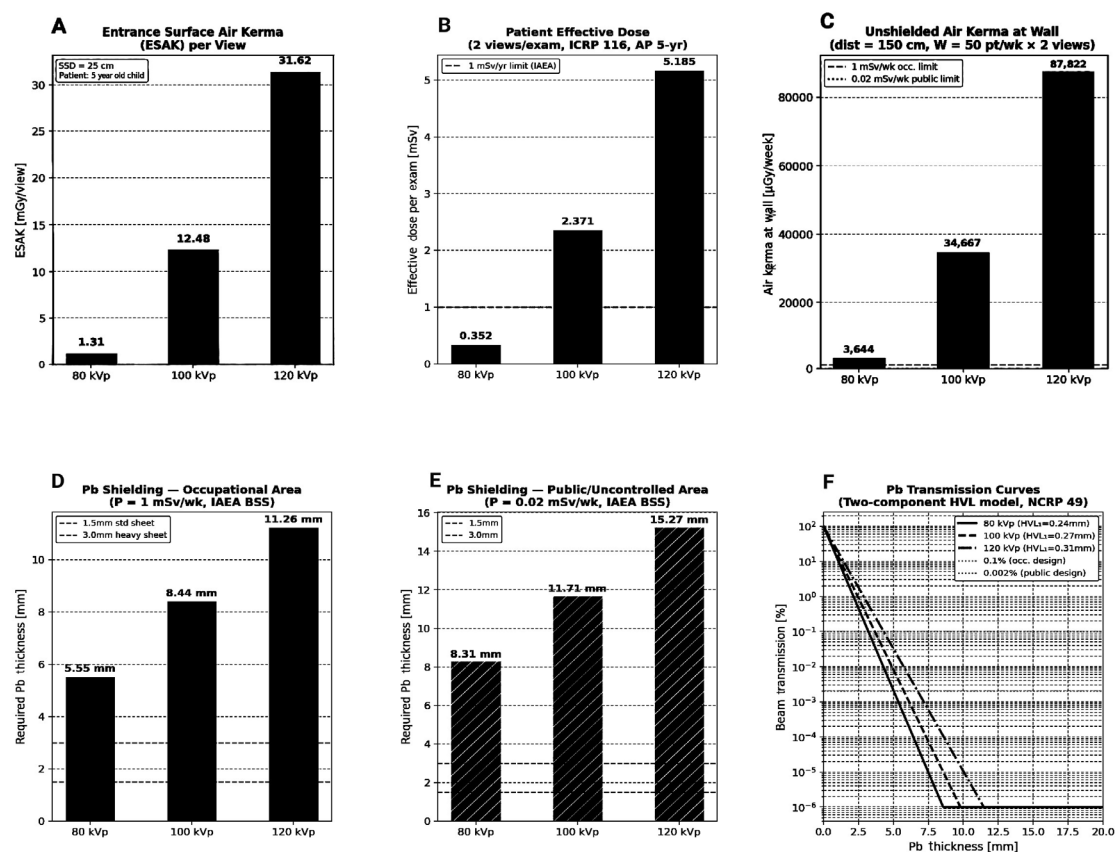


Figure 4. Shielding design results. (A) ESAK per view across tube voltages. (B) patient effective dose per examination. (C) Unshielded weekly air kerma at the 1.5 m wall. (D) Required Pb thickness for occupational and public areas. (E) Pb transmission curves showing the two HVL model. Workload: 50 patients/week $\times$ 2 views. Abbreviations: ESAK: Entrance surface air kerma; HVL: Half-value layer; SSD: Source-to-surface distance.

shielding thicknesses (Table 4) assume conservative worst-case occupancy and use factors ($T = U = 1$) and NCRP 49 Pb HVL data; facility-specific T and U factors and, where applicable, more current shielding guidance such as NCRP 151¹⁹ should be applied by a qualified radiation protection expert before these figures are used for room design. Finally, because the simulated geometry (1 mm focal spot, SSD = 25 cm, Hitachi MBR1618RBE cabinet irradiator) corresponds to a small animal/preclinical research unit rather than a clinical diagnostic radiography room, the dosimetric and shielding results in Sections 3.4 and 3.5 characterize this specific research irradiator; they are presented as an illustrative downstream application of the simulated spectra and should not be read as clinically applicable pediatric diagnostic radiology reference values without geometry-specific revalidation.

4. Conclusion

This study demonstrates that PHITS 3.35 with EGS5 physics reproduces the expected diagnostic X-ray spectral shape for a cabinet irradiator at 80, 100, and 120 kVp, capturing both the bremsstrahlung continuum and tungsten K-characteristic emission lines, with mean photon energies of 57.9–67.6 keV consistent with published analytical models. Corrected HVL₁ values of 3.08–3.43 mm Al, derived analytically from the simulated spectra using NIST XCOM aluminum attenuation data, fall within the range reported for similarly filtered diagnostic beams in the literature (an order-of-magnitude units error present in the originally submitted HVL calculation has been corrected in this revision), and homogeneity coefficients systematically below unity confirm polychromatic beam character at all voltages, with the degree of polychromaticity decreasing at higher kVp. Analytically derived patient effective dose for a 5-year-old pediatric phantom at this irradiator's short SSD of 25 cm was 0.35–5.19 mSv per examination; because these values use a short SSD, simplified collimation source geometry, and an effective dose quantity not directly comparable to published entrance-dose-based DRLs, they should not be interpreted as evidence that clinical pediatric DRLs are or are not exceeded. Lead shielding requirements of 8.3–15.3 mm Pb for public areas and 5.6–11.3 mm Pb for occupational areas at a 1.5 m wall distance represent conservative, worst-case ($T = U = 1$) bounding estimates; realistic facility-specific occupancy and use factors would reduce these values, but the reduced thicknesses were not recalculated here and should be determined by a qualified radiation protection expert following current guidance (e.g., NCRP 151) before use in facility design. A practical PHITS implementation detail, the distinction between material code 0 for interior vacuum and code 1

exclusively for the outer void boundary, is documented to help other users adapting EGSnc-based models to PHITS. No independent experimental or code-to-code validation was performed in this study. This work is best characterized as a PHITS-based modeling exercise for a specific cabinet X-ray irradiator, illustrating a reproducible spectral simulation, dosimetry, and shielding estimation workflow; extension of these results to clinical pediatric diagnostic radiology facilities would require validation against measurement or an independent spectrum code, projection-specific dose coefficients, and facility-specific shielding recalculation.

Acknowledgments

The authors acknowledge the JAEA PHITS development team for the provision of PHITS version 3.35 and associated documentation.

Funding.

None.

Conflict of interest

The author declares no conflict of interest.

Author contributions

This is a single-authored article.

Ethics approval and consent to participate

Not applicable. This study involved only Monte Carlo simulation and analytical dose/shielding calculations using published reference data and coefficients; no human participants, patient data, biological samples, or animal subjects were involved.

Consent for publication

Not applicable.

Availability of data

PHITS input files, spectral data, and analysis scripts are available from the corresponding authors upon reasonable request.

Further disclosure

During preparation of this revised manuscript, the author used a generative AI writing assistant to check the consistency of corrected numerical values across the text, tables, abstract, and conclusion. The author reviewed, verified, and edited all AI-assisted content, is responsible for the scientific accuracy of the corrected calculations, and takes full responsibility for the published article.

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