

Bowtie and Spectral Filtration in CT for Radiographic Practice

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computed tomography; bowtie filter; patient centring; spectral filtration; radiotherapy

ABSTRACT

This review connects CT bowtie and spectral filtration with radiographic practice through a common framework linking patient geometry, exposure control, dose endpoints and task-specific image performance. A targeted narrative synthesis combined web and Europe PMC searches with reference follow-up. Evidence was compared by mechanism, study design, comparator and endpoint, without pooling or formal risk-of-bias assessment. The synthesis identified two interacting pathways through which off-centring affects exposure: mismatch between patient geometry and the bowtie-shaped beam, and changes in localiser information used by automatic exposure control. Scanner and endpoint differences prevent a universal off-centring penalty. Camera assistance can improve positioning or workflow without consistent downstream dose benefits. Spectral filtration changes attenuation and dose-conversion relationships, requiring task-specific assessment. Photon-counting studies identify additional calibration and scatter-correction requirements; adaptive attenuators remain a research perspective. Centring should be judged against the intended anatomical target and scanner protocol. Positioning accuracy, scanner dose indices, organ dose and diagnostic performance are distinct outcomes that do not necessarily improve together. Interpreting them together offers a more defensible basis for practice than applying a universal off-centring penalty. Radiographers should assess camera recommendations against the intended anatomy and localiser geometry. Protocol changes affecting quantitative CT require joint validation with radiologists and medical physicists. Local audits should examine positioning, dose and image quality separately.

INTRODUCTION

Beam-shaping filtration is among the least visible components of CT optimisation to the radiographer. The filter is embedded within scanner hardware and is usually selected through protocol and scanner logic rather than direct manual control. Yet it operates on a geometry established at the patient table. Its effectiveness therefore depends on practical decisions about patient position, acquisition field and protocol [1,2].

Reviews already cover patient positioning, automatic centring and tin filtration [3,4,5]. The unresolved practical issue is how to interpret a change in centring when geometric alignment, localiser-based exposure control and the chosen dose endpoint do not necessarily improve together. This review addresses that question by linking mechanisms to study design and task-specific checks. Its contribution is an interpretive framework for radiographic practice, rather than a pooled effect estimate or a new positioning algorithm.

MATERIALS AND METHODS

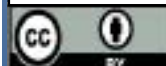
This narrative review used topic-led web searches, a focused Europe PMC update on 6 September 2026, and targeted web and DOI-based checks on 7 September 2026. Sources were also identified from

author-supplied publications and reference lists. Selection was guided by relevance to bowtie geometry, patient positioning, exposure control, spectral filtration and task-specific imaging outcomes. The initial search was unrecorded and had no predefined eligibility criteria. Scopus and Web of Science were not searched; coverage was not intended to be comprehensive.

Evidence was synthesised by mechanism, study design, comparator and endpoint, without pooling or formal risk-of-bias assessment. Phantom measurements, simulations and clinical comparisons were interpreted separately. Claims were checked against abstracts and accessible full-text sections; incomplete full-text access limited methodological appraisal. A retrospective evidence map summarises selected pivotal studies, their comparators and the limits of the inferences drawn (Supplementary Table S1). This map does not reconstruct the unavailable original search log or establish systematic coverage.

RESULTS

This section presents the findings of the narrative synthesis, organised by physical mechanism and imaging application. It reports evidence from the cited literature rather than new experimental or patient data.



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Table 1. Distinct outcomes in centring and filtration studies

Outcome	What it measures	What it does not establish
Positioning accuracy	Deviation from isocentre or from the protocol-defined anatomical target	Dose delivered or diagnostic performance
CTDIvol and DLP	Scanner output referenced to a standard phantom geometry	Organ dose in an individual patient, particularly across different spectra
Organ dose	Absorbed dose in a specified organ, by measurement or modelling	Diagnostic adequacy for the clinical task
Image noise	Variability of attenuation values in a uniform region	Lesion detection or quantitative accuracy
Task-specific diagnostic performance	Detection or quantification for a prespecified task	Equivalence, unless an acceptance margin was defined in advance

Outcomes reported in the reviewed literature. Improvement in one outcome does not imply improvement in another, and a non-significant difference does not establish equivalence. This table defines endpoints; it does not rank study quality or provide conversion factors between outcomes.

3.1. Physical principles of bowtie filtration

Across a CT projection, attenuation varies markedly. Central rays through the torso may cross several tens of centimetres of tissue, whereas peripheral rays encounter shorter paths or may miss the patient. Without shaping, peripheral detector channels can receive far more photons than central channels, increasing detector dynamic-range requirements and

directing unnecessary fluence through thin peripheral regions [1,2].

A bowtie is a shaped attenuator whose thickness generally increases with fan angle. By reducing peripheral fluence before it reaches the patient, it better matches incident photon statistics to object geometry, reduces the dynamic range of incident fluence presented to the detector and improves spatial dose efficiency [1,2]. Figure 1 illustrates this geometric correspondence. The filter also changes scatter conditions: reducing incident fluence can reduce scatter generated in the patient, while the filter itself can generate scattered radiation that requires correction [6].

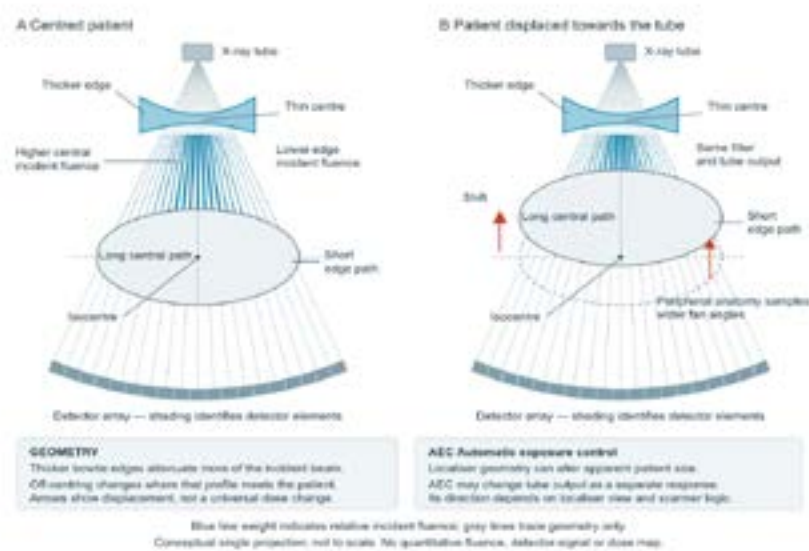


Figure 1. Why bowtie filtration depends on patient centring. In a centred patient (A), the thinner filter centre admits greater incident fluence along longer tissue paths, while thicker filter edges reduce fluence towards shorter peripheral paths. With displacement towards the tube (B), peripheral anatomy samples wider fan angles; the dashed contour shows the original position. Blue line weight represents relative incident fluence before patient attenuation; grey rays show geometry only. Detector shading identifies the detector elements, not signal or dose. Arrows indicate anatomical displacement. The boxes distinguish geometric mismatch from localiser-dependent automatic exposure control (AEC) [1,2,3,7,8,9,10,11]. This is a conceptual single projection, not a quantitative dose map or a scale drawing.





3.1.1. Filter material and spectral effects

Bowtie performance depends on material as well as contour. Rays crossing thicker peripheral portions undergo greater attenuation and beam hardening than central rays, so the effective spectrum varies with fan angle. Calibration and correction algorithms must account for this spatial dependence. Geometric beam shaping and deliberate spectral shaping should therefore be distinguished: the conventional bowtie primarily redistributes fluence across angle, whereas added filters such as tin are intended to alter the energy distribution of the beam [1,2].

3.1.2. Bowtie profiles and scanner-specific allocation

“The bowtie” is a convenient singular for profiles that differ among manufacturers and scanner generations. Direct transmission measurements show that some platforms use several filters allocated by patient size or examination type, whereas others apply a single measured profile across a broader set of modes [1,2]. Bowtie design can vary independently in angular contour and central baseline attenuation: filters may share similar central attenuation while differing peripherally, or share a similar contour while differing in central thickness [2]. This distinction matters because filter count alone does not describe how a system manages patient size and task-specific fluence. Filter labels often refer to examination mode or patient-size categories, but do not establish a universal ranking of angular shaping or central attenuation [1,2]. The measured profiles in Table 2 illustrate why filter names cannot substitute for scanner-specific allocation guidance. The distinction between contour and central attenuation also limits the transfer of findings from one platform to another.

Allocations and measured characteristics are those reported by Yang et al. using direct transmission measurements and expressed as aluminium-equivalent thickness [2]. Filter nomenclature and allocation vary between scanner generations and manufacturers. These historical measurements are not a current manufacturer catalogue or protocol-selection guide. Patient-specific dose modelling also requires validation of the source description. Rosendahl et al. derived equivalent CT source models non-invasively and compared Monte Carlo calculations with measurements at five locations in physical phantoms on two scanners [12]. Bowtie geometry and the tube starting angle contributed to uncertainty; agreement at sampled locations does not establish accuracy for every organ in a patient. This provides a methodological link between scanner-specific filtration and personalised dosimetry, rather than a dose-reduction estimate.

3.1.3. Acquisition field and filter selection

On systems that link acquisition field of view to bowtie selection, changing the acquisition field may also change the filter profile. This acquisition setting should be distinguished from the reconstructed display field of view: cropping a reconstructed image does not retrospectively change beam filtration. Acquisition-field selection should follow scanner-specific guidance and provide appropriate anatomical coverage for the clinical task [2].

3.2. Patient centring and dose distribution

Bowtie performance depends on alignment between patient anatomy and the fluence profile produced by the selected filter. When the patient is displaced from isocentre, the same angular attenuation profile is ap-

Table 2. Bowtie profiles reported for three CT platforms in a 2019 measurement study

Platform	Profiles measured	Mode allocation or measured relationship	Design implication
GE Revolution	3	Large: large body and cardiac large; Medium: head, medium body and cardiac medium; Small: small head, small body, cardiac small, paediatric head and paediatric body	A limited set of profiles is shared across several anatomical and acquisition categories
Toshiba/ Canon Aquilion ONE VISION	3	Large and medium share aluminium-equivalent thickness at the centre but differ in profile; medium and small share profile shape but differ in central thickness	Contour and baseline attenuation are treated as partly independent design variables
Philips iQon	1	One measured profile across infant and adult head, adult body, cardiac and infant body modes	A single measured profile is applied across a broad range of anatomical and acquisition categories



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plied to different geometry, changing surface dose, attenuation, detector fluence and noise. The mismatch is geometric rather than a simple change in total output: a patient displaced towards one side of the beam encounters a different part of the attenuation profile, while the distribution of transmitted fluence across detector channels also changes [3,7,8,9].

Phantom studies demonstrate that off-centring can increase surface dose and image noise, but their results describe the tested object, scanner and acquisition settings [7,8]. Toth et al., for example, reported surface-dose increases of 18% and 41% for positions 3 and 6 cm below isocentre in a 32-cm CTDI phantom, with corresponding noise increases of 6% and 22% [7]. These values establish the importance of geometry under those conditions; they do not predict an individual patient’s organ-dose change.

The direction of a local dose change need not match the change in total scanner output. Ou Hadda et al. measured lower peripheral dose on the side towards which a phantom was displaced and higher dose at opposite measurement points [9]. This does not contradict reports of increased surface dose under other configurations: the endpoint, measurement location and exposure-control response differ. A broader review likewise found substantial dependence on scanner, patient size, localiser orientation and bowtie configuration [3]. Acceptable reconstructed appearance cannot establish optimal dose distribution, and a lower dose at one location does not establish an overall patient benefit.

A recent fixed-output experiment further separated local redistribution from the weighted dose index. Oustous et al. measured dose at standard locations in a 32-cm PMMA phantom and noise in a separate Catphan experiment. Local point-dose changes reached approximately 48%, while CTDI_w stayed within ±5%; iterative reconstruction reduced noise but retained a positioning penalty [13]. The large offsets and single-scanner design limit transferability. These measurements concern phantom locations, not patient organ doses, and do not quantify an AEC response.

Interpretive synthesis of study conditions and endpoints. Study design identifies the question addressed, not a quality score. This table provides neither pooled estimates nor patient-specific correction factors.

3.2.1. Prevalence and interpretation of off-centring

In 57,621 adult chest, abdominal and pelvic CT acquisitions, 83.7% were performed with the patient centroid below gantry isocentre; mean displacement was 1.7 cm, with some examinations more than 12 cm below isocentre [20].

A 2026 cardiac CT study of 399 adults reported off-centring greater than 10 mm in 83% of examinations anteroposteriorly and 50% laterally; 39% exceeded 30 mm anteroposteriorly. The institutional protocol centred on the thorax rather than the heart, so these deviations did not necessarily represent radiographer positioning errors. Off-centring was not independently associated with volume CT dose index (CTDI_{vol}) or dose-length product (DLP) after adjustment. Scenario modelling suggested a hypothetical reduction of about 12% with ideal centring; this was not an observed intervention effect [16]. The findings illustrate the importance of distinguishing the intended anatomical target, the positioning protocol and the reference used to define off-centring.

3.2.2. Localiser geometry and automatic exposure control

Off-centring acts through two pathways that must be distinguished. First, the bowtie-shaped beam encounters a different patient geometry. Second, localiser magnification or minification can change the information used by automatic exposure control (AEC). The second pathway depends on scout direction and vendor implementation, including tube-current and tube-voltage selection [10,11]. Consequently, a fixed-output phantom experiment and

Table 3. Why reported off-centring effects differ

Study feature	Interpretive consequence	Evidence
Fixed exposure versus AEC	Localiser-driven output changes can accompany geometric mismatch.	Scout direction and exposure-control studies [10,11].
Surface, peripheral or organ dose	A local decrease can coexist with an increase elsewhere; endpoints are not interchangeable.	Phantom, postmortem and Monte Carlo studies [7,8,9,14,15].
Body centroid versus anatomical target	Deviation may reflect protocol design rather than operator error.	Cardiac CT observational study [16].
Camera accuracy versus dose benefit	More accurate positioning does not imply a fixed dose reduction; reconstruction differences can confound image-quality comparisons.	Meta-analysis and clinical camera comparisons [4,17,18,19].
Weighted index versus local dose	A stable CTDI _w can coexist with substantial peripheral redistribution.	Fixed-output phantom measurements; not patient organ dosimetry [13].



an AEC-enabled clinical study need not produce the same direction or magnitude of dose change. Protocol comparisons should identify localiser orientation and AEC settings rather than attributing every difference to the bowtie alone. A head-phantom study found opposing dose-index trends with anteroposterior and posteroanterior localisers, while vertical displacement did not alter dose indices with the lateral localiser [21]. This supports reporting localiser direction when interpreting a centring experiment; it does not establish a universal preference for lateral localisers.

3.2.3. Camera-assisted patient positioning

A meta-analysis of nine comparative studies reported smaller positioning offsets with automatic systems and dose reductions ranging from 5.7% to 31% across studies [4]. The range is not a pooled estimate of the benefit achievable on any scanner. Positioning accuracy is a more direct endpoint than patient dose, which also depends on patient mix, acquisition settings and exposure control. The review

reported mixed image-quality findings, supporting a cautious interpretation of downstream benefits. Dane et al. reported improved isocentre accuracy with camera assistance [17]. Camera proposals should still be checked against the intended anatomical target and local protocol. However, Nelson et al. found improved positioning without a significant dose change after accounting for water-equivalent diameter [18]. In a prospective study of 596 participants, Ito et al. reported shorter room time with camera assistance but no significant differences in offset, CTDIvol, DLP or background noise [19]. These findings support separate assessment of workflow, positioning and dosimetry. In a prospective CT kidney, ureter and bladder comparison of 286 patients, Kundu et al. reported smaller offsets, lower dose indices and better image-quality measures with automatic positioning. However, automatic and manual groups used deep-learning and iterative reconstruction, respectively, so image-quality differences cannot be attributed to positioning alone. Only patients with normal BMI were included [22].

Table 4. Evidence-linked considerations for local protocol review

Situation	Consideration for review	Reason	Boundary and evidence
Camera proposal differs from expected position	Check detected contour and intended target against the local protocol.	Surface detection and the clinical target may differ.	Use professional assessment; no universal override rule [4,16,17].
Position changes after the localiser	Check the local workflow for AEC planning before acquisition.	AEC may use information from the earlier geometry.	Do not routinely repeat a localiser without clinical and protocol justification [10,11].
Changing acquisition field or patient-size mode	Confirm anatomical coverage and scanner-specific filter allocation.	Mode labels can map to different or shared profiles.	Reconstruction cropping does not alter delivered filtration [1,2].
Introducing spectral filtration in quantitative CT	Seek task-specific protocol and quantitative-measurement validation.	Spectral and reconstruction changes can alter measured attenuation.	Neither a calcium threshold nor correlated cyst scores establishes interchangeability [5,23,24,25].
Filtration change in proton planning CT	Refer calibration implications to the medical physics team.	Stopping-power estimation depends on CT-number calibration.	Retain validated treatment-position and calibration workflows [26,27].
Considering adaptive-filter claims	Separate simulation endpoints from clinical evidence.	Benefits depend on detector and optimisation assumptions.	No clinical count-rate intervention follows directly from these simulations [28,29,30,31,32].
Comparing dose reductions across spectra	Identify the dose endpoint and use spectrum-appropriate conversion or modelling.	Organ dose per unit CTDIvol can change with beam quality.	CTDIvol reduction is not an equal organ-dose reduction [33,34].



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These considerations support local protocol review within established clinical and medical-physics governance. They are not validated decision rules or independent instructions to change scanner settings.

3.3. Spectral filtration and task-specific image quality

The bowtie primarily shapes the angular distribution of incident fluence. Spectral filtration alters the energy distribution of the beam and can change quantitative attenuation measurements. Tin, with a K-edge near 29 keV, preferentially attenuates lower-energy photons that contribute disproportionately to absorbed dose in many examinations while having limited probability of traversing thicker anatomy. The resulting beam has a harder spectrum as well as reduced photon fluence [5,23]. Figure 2 separates spatial fluence shaping from spectral modification.

the same percentage reduction in organ dose. This finding does not imply that an optimised tin-filtered protocol increases absolute organ dose; it changes how the reported reduction must be interpreted.

A 2025 systematic review of adult non-contrast CT evaluated tin filtration combined with iterative reconstruction. Dose reductions were reported, but increased image noise was common and most comparisons did not use a consistent preset noise level. These results concern combined optimisation protocols and do not isolate the effect of tin filtration [5]. Lower CTDIvol alone does not establish equivalent diagnostic performance. In a coronary-calcium phantom study, tin-filtered acquisition at 100 kVp (Sn100) reduced calcification detectability and Agatston scores; lowering the threshold from 130 to 117 HU improved comparability with 120 kVp within

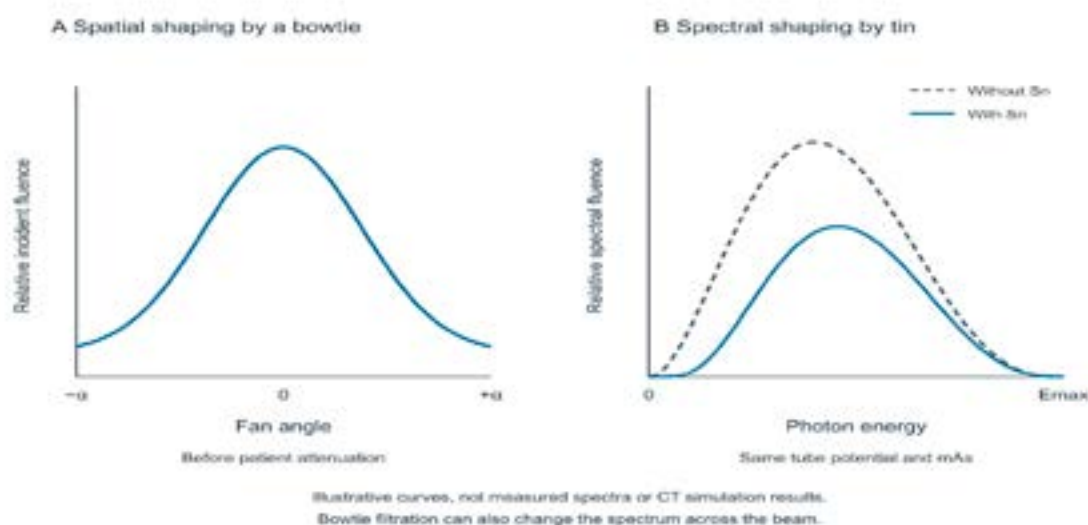


Figure 2. Spatial and spectral beam shaping. Panel A shows an illustrative angular fluence profile before patient attenuation. Panel B illustrates preferential attenuation of lower-energy photons by tin at the same tube potential and mAs, with a common maximum energy. The curves are mathematical illustrations, not measured spectra or outputs of a validated CT simulation. Bowtie filtration can also change the spectrum across the beam [1,2,23].

Greffier et al. measured half-value layer and calculated mean weighted energy for standard and tin-filtered beams at selected tube potentials. At the same tube voltage, tin filtration increased both measures of beam quality and changed CT numbers for bone and acrylic inserts. Task-based phantom assessment showed benefits that depended on the lesion and acquisition protocol [23]. These results support evaluating spectral filtration against the diagnostic task rather than treating it as a uniform dose-reduction factor.

Dose-index conversion introduces a separate spectral issue. In Monte Carlo models validated against scanner CTDIvol measurements, Fong et al. evaluated 53 computational subjects and found higher CT-_{DIvol}-normalised organ-dose and DLP-normalised effective-dose coefficients for harder, tin-filtered spectra [33]. Consequently, equal CTDIvol need not imply equal organ dose across spectra, and a percentage reduction in CTDIvol should not be reported as

that experiment [24]. This is a study-specific adjustment, not a validated universal threshold for Sn100 protocols. A prospective intraindividual study of 201 participants subsequently reported high agreement in calcium-risk categories using tin-filtered research acquisitions and reconstruction designed to reproduce 120-kV attenuation [35]. A retrospective study of 30 patients combined tin filtration, high-pitch acquisition and denoising. Total calcium scores correlated with the standard protocol, but vessel-level differences remained; correlation and non-significant total-score differences do not establish interchangeability or an isolated filtration effect [36]. In paediatric sinus CT, Sn100 reduced CTDIvol in phantom and patient comparisons. Lower eye entrance dose was measured in the phantom; blinded patient image-quality scores did not differ significantly between protocols. A non-significant difference does not establish statistical equivalence [37]. Protocol transfer therefore requires task-specific



validation, including quantitative attenuation where relevant.

Clinical evidence also illustrates why the imaging task matters. Fong et al. combined phantom optimisation with a 100-patient colonography comparison, assigning conventional and tin-filtered protocols to different patient positions and using seven blinded readers. CTDIvol decreased by 37%, whereas estimated effective dose decreased by 29%. Colon image-quality ratings were usually comparable or better with tin filtration, but this occurred less consistently for extracolonic findings [34]. The comparison changed tube potential as well as filtration, and subjective image-quality ratings do not establish equivalent lesion-detection sensitivity. Benefits therefore apply to the tested protocol and task, not to tin as an isolated intervention.

Evidence for spectral filtration also extends to silver. Oshima et al. compared silver and copper filters across dose levels and three reconstruction methods in a chest phantom, assessing both image-quality metrics and nodule detection by receiver operating characteristic analysis [38]. Results favoured silver and deep-learning reconstruction under the tested conditions. Comparisons using the same reconstruction help distinguish filtration from reconstruction effects, but phantom detection performance does not establish screening effectiveness in patients or superiority over tin.

In a prospective single-centre study, Golbus et al. obtained standard and ultra-low-dose chest CT in 60 women with lymphangioleiomyomatosis [25]. Silver filtration with deep-learning reconstruction reduced estimated effective dose by 85.5%, while cyst scores remained strongly correlated with the standard protocol. Median scores nevertheless differed by 1.5 percentage points, with negative bias in the low-dose comparison. Acquisition settings and reconstruction changed alongside filtration, precluding attribution of the reduction to silver alone. Effective dose was calculated from DLP using the same conversion coefficient for both spectra, not measured as organ dose. These findings support task-specific quantitative evaluation; they do not establish interchangeable cyst scores or general diagnostic equivalence.

3.4. Filtration in radiotherapy CT simulation

In CT simulation for radiotherapy, the bowtie shapes the incident kilovoltage beam by attenuating peripheral rays more strongly than central rays. This angular fluence profile is designed to match the shorter peripheral and longer central paths through the patient, improving dose efficiency and reducing the range of photon fluence reaching the detector. Its performance therefore depends on patient geometry and alignment with the selected filter profile [1,2].

A key requirement in treatment planning is compa-

tibility with the commissioned CT-number calibration used for dose calculation. Photon planning uses a validated conversion from CT numbers to the density quantities required by the dose algorithm; proton planning depends on the CT-number to stopping-power relationship [26,27,39]. Tube potential, filtration and reconstruction settings must therefore be assessed within the local calibration workflow. This does not prohibit protocol variation: it requires validation of the settings and calibration used together. Similar calibration requirements apply in quantitative diagnostic CT [23,24].

Treatment-position reproducibility and immobilisation may limit the anatomical centring achievable. Flat couch tops and immobilisation devices can affect the position achievable within the gantry; their presence alone does not demonstrate bowtie mismatch. The practical implication is to assess patient-bowtie alignment and localiser geometry within the agreed simulation protocol, while preserving the required treatment position. Diagnostic centring targets should not be transferred automatically to a radiotherapy setup.

In research on spectral cone-beam CT for proton imaging, Leibold et al. compared six configurations using a theoretical lower bound on stopping-power-ratio variance [40]. The model links spectral separation, source operation and detector response; it does not establish achieved proton-range accuracy or a validated treatment-planning workflow.

3.5. Photon-counting CT calibration and detector count-rate capability

Photon-counting CT extends the physical discussion to incident photon flux relative to detector count-rate capability. At sufficiently high flux, finite processing times can cause pulse overlap, count losses and degraded energy discrimination [41]. This is a limitation of detector performance; it does not establish a new console setting or independently validated radiographer intervention.

Estimates based on 1,980 clinical CT examinations showed substantial variation in incident photon count rates with patient attenuation, technique, filtration and geometry, exceeding 1×10^9 counts/mm²/s in simulations at 1000 mA [41]. These are estimates of incident rates, not direct measurements of recorded counts in a clinical photon-counting detector. Regions in which rays traverse little tissue or miss the patient are especially relevant to source-side fluence shaping.

Count-rate capability is not the only consideration for filtration in photon-counting CT. Zhang et al. used water-phantom simulations and benchtop experiments to show object-size- and spectrum-dependent non-uniformity with bowtie filtration. Beam-hardening correction mitigated these effects when dedicated calibration parameters were used [42]. Hennemann et al. subsequently investigated



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joint correction of patient and bowtie scatter using Monte Carlo data and anthropomorphic phantom measurements on a clinical photon-counting scanner [6]. Spectral information improved correction, particularly in virtual monoenergetic images. Validation covered a standard bowtie and forward scatter; other filter geometries and dual-source cross-scatter remained outside that evaluation. These studies support coordinated filtration and correction design, without demonstrating improved clinical diagnostic outcomes.

Research-stage dynamic attenuators aim to adapt to changes in patient attenuation across projection angle and longitudinal position, balancing dose, noise and detector loading [28,29,30,31,32]. Their performance should be compared at a specified dose or image-quality target, rather than by count-rate reduction alone.

A fluid-filled attenuator simulation reported substantial count-rate reductions at matched signal-to-noise ratio [30]. Other models evaluated dose efficiency or spectral-image precision under different detector and optimisation assumptions [28,29,31,32].

Lin et al. compared six simulated cone-beam CT protocols at matched image noise, finding favourable dose distributions and noise-power-spectrum isotropy with a conceptual fluid dynamic bowtie [43]. The prominently reported reductions used a baseline without attenuation filtration or current modulation; whole-region averages also differed from individual organ endpoints. They therefore do not quantify savings over an optimised static-bowtie protocol. Patient-specific geometry and actuation remained to be developed.

Adaptive filtration would require reliable actuation, calibration and compatible reconstruction. These simulation results establish technical possibilities, but their translation to commercial systems and patient outcomes remains unproven. They do not define a radiographer-controlled count-rate intervention [28,29,30,31,32,43].

DISCUSSION

4.1. Interpretation across mechanisms and endpoints

The reviewed evidence supports interpreting positioning, exposure control and image performance as related but distinct outcomes. Geometric alignment can improve without a measurable change in scanner dose indices, while spectral changes can alter the relationship between those indices and patient-dose estimates. Apparent differences between studies therefore need to be considered alongside their comparators, acquisition settings and measured endpoints. The practical value of this synthesis lies in making these distinctions explicit when reviewing local protocols.

4.2. Implications for radiographic practice

Table 4 supports local protocol review by linking practical situations to mechanisms, evidence and interpretive limits; it does not establish intervention thresholds.

A local audit could compare deviation from the protocol-defined target with dose indices and task-relevant image quality, accounting for scanner, localiser orientation, patient size and acquisition settings. These outcomes should be reported separately. When organ dose is the question, spectrum-appropriate measurements or modelling are needed; CTDIvol and DLP alone cannot answer it [3,4,10,11,14,15,33]. This proposed audit application has not been evaluated by the review.

For quantitative imaging and radiotherapy planning, filtration changes require joint review of acquisition settings and calibration with the radiologist and medical physicist. Validation should use the intended measurement or diagnostic task; research thresholds should not be transferred directly between protocols [24,26,27,39].

4.3. Limitations of the review

Topic-led selection and narrow update queries may introduce selection bias; the absence of a complete search log prevents assessment of search completeness. Some studies were assessed only through abstracts, limiting evaluation of methods and confounding. No formal risk-of-bias assessment or certainty grading was performed; Supplementary Table S1 identifies interpretive limitations rather than quality scores. Phantom measurements, simulations and observational comparisons answer different questions, and vendor-specific results may not transfer between systems. Industry involvement and proprietary scanner details in some studies warrant particular attention to independent validation. The proposed practice considerations have not been tested prospectively.

4.4. Future research

Future studies should report positioning, scanner indices, organ-dose estimates and task-specific diagnostic performance as separate endpoints, with controlled reconstruction and explicit localiser geometry. Imaging equivalence requires a prespecified task and acceptance margin rather than a non-significant difference alone. Radiotherapy studies should assess centring within reproducible treatment setups and validated calibration workflows. Photon-counting evaluations should report filter geometry and correction conditions alongside count-rate or image-quality results. Prospective evaluation is needed before the proposed practice considerations can be treated as a decision tool.

CONCLUSIONS

Bowtie and spectral filtration influence different but coupled parts of CT optimisation. Off-centring has no single transferable dose penalty, and a change in CTDIvol does not establish the same change in organ dose or diagnostic performance. Interpretation requires the anatomical target, localiser view, exposure-control settings, spectrum, reconstruction and measured endpoint. Camera assistance can improve positioning or workflow, with variable downstream dose effects.

For radiographers, the practical contribution is to distinguish a positioning discrepancy from a protocol or calibration problem. Quantitative imaging and radiotherapy planning require validation of acquisition and calibration together. Photon-counting studies extend this requirement to spectral and scatter corrections; adaptive attenuation remains a research perspective within the evidence considered.

Declaration of generative AI and AI-assisted technologies in the manuscript preparation process

During the preparation of this work the authors

used ChatGPT and Codex (OpenAI) for language editing and for code development for the conceptual figures, and Claude (Anthropic) for document formatting, for consistency checks of citations and reported values against the reference list, and for editorial revision of the supplementary material. The figures contain no patient images or measured data. After using these tools, the authors reviewed and edited the content as required and take full responsibility for the content of the publication.

Supplementary materials

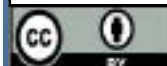
Supplementary Table S1 provides a retrospective evidence map of selected pivotal studies. Separate figure files and legends are also provided. The original search log is unavailable.

Ethics and data availability

This narrative review reports no new human or animal participant data. No new patient datasets were generated. The cited publications constitute the evidence base; limitations of source identification and access are described in the methods and limitations sections.

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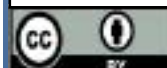
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Author contributions

All authors contributed to this work and have read and approved the manuscript.

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