INDIAN JOURNAL OF ALLIED HEALTH SCIENCE (IJAHS)
Volume 2, Issue 3, Jul-Sep 2026

Online ISSN: 3108-3005

INDIAN JOURNAL OF ALLIED HEALTH SCIENCE (IJAHS)
www.ijahs.org

Review Article

Contemporary Strategies for Radiation Dose Optimization in Computed Tomography: Balancing Diagnostic Image Quality and Radiation Exposure

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Received: 28/04/2026
Accepted: 26/08/2026
Published: 13/09/2026

DOI:10.66159/IJAHS.2026.2309

ABSTRACT

Background: Computed tomography (CT) is an essential imaging modality because of its rapid data acquisition, high spatial resolution, and broad clinical applications. However, increasing utilization has raised concerns regarding radiation exposure and its potential risks. Radiation dose optimization is therefore essential to achieve diagnostically adequate image quality while minimizing patient exposure. This review critically synthesizes current strategies for optimizing radiation dose in CT, focusing on acquisition parameters, patient-size-based dose adjustment, image reconstruction, artificial intelligence (AI), photon-counting CT (PCCT), pediatric imaging, image quality, and dose quantification.

Methodology: A literature review was conducted using PubMed, Scopus, ScienceDirect, and Google Scholar, covering publications from January 2010 to August 2026. Of 300 identified articles, 42 duplicates were removed. Following title and abstract screening, 258 articles remained, of which 84 underwent full-text assessment. Ultimately, 70 articles were included in the review.

Conclusion: CT radiation dose optimization requires protocol optimization, appropriate patient-size adjustment, advanced reconstruction techniques, and clinical awareness. AI-based reconstruction and PCCT offer promising opportunities to improve dose efficiency while maintaining diagnostically acceptable image quality.

Keywords: Computed Tomography; Radiation Dose Optimization; Image Quality; Artificial Intelligence; Image Reconstruction; Photon-Counting CT; Pediatric CT; Radiation Exposure.

INTRODUCTION

In recent years the role of Computed Tomography (CT) is widely used in emergency medicine, oncology, neurology, cardiology, trauma, and in several other medical applications lies in the fact that it offers fast volumetric imaging along with detailed anatomy. However, its increasing clinical use has intensified the need for appropriate justification and optimization of radiation exposure [1, 2, 3, 4]. The fundamental objective is not simply to achieve the lowest possible numerical radiation dose, but to use the lowest dose that provides image quality adequate for the specific diagnostic task [1]. Collaborative protocol development involving radiologists, medical physicists, and radiologic technologists [1, 2]. CT dose optimization is therefore a multidimensional process that includes appropriate examination justification, patient-specific acquisition, optimization of scan coverage, selection of tube potential and tube current, use of automatic exposure control, appropriate reconstruction techniques, and continuous assessment of diagnostic image quality [1, 5-7]. Diagnostic reference levels (DRLs) can help identify unusually high or variable radiation dose practices; however, they are optimization tools rather than dose limits and should not be considered target doses for individual patients [8, 9]. Recent technological developments have further expanded the possibilities for CT dose optimization. Hybrid and model-based iterative reconstruction techniques can reduce image noise compared with filtered back projection (FBP) [10-15], while deep learning reconstruction (DLR) has the potential to improve noise reduction and preserve diagnostically useful image texture at lower radiation doses [16, 17]. Photon-counting CT (PCCT) introduces direct photon detection, smaller detector elements, spectral imaging capabilities, and potentially improved dose efficiency [18]. These technologies are promising, but their clinical value depends on the imaging task, patient size, protocol, reconstruction strength, and diagnostic endpoint [17, 19, 20].

METHODOLOGY

This study was conducted as a review to present a critical analysis of the existing approaches to dose reduction in CT scanning from a clinical point of view. This review was conducted because the available literature has different variations including differences in CT technologies, clinical indications, patient populations, scanning protocols, radiation-dose measurements, reconstruction techniques, and image-quality assessment methods.

A literature search was conducted using studies published in PubMed, Scopus, ScienceDirect, and Google Scholar from January 2010 to August 2026. Earlier landmark studies and authoritative professional guidance were included when necessary to describe the established CT dose metrics, radiation protection principles, and foundational optimization concepts. Search terms were combined using Boolean operators and included "computed tomography" or "CT", "radiation dose", "dose optimization", "dose reduction", "CTDIvol", "DLP", "SSDE" (Size-specific dose estimate), "automatic exposure control", "tube current modulation", "tube voltage", "iterative reconstruction", "model-based iterative reconstruction", "deep learning reconstruction", "artificial intelligence", "photon-counting CT", "paediatric CT", "diagnostic reference levels", "image quality", "low-contrast detectability", and "diagnostic performance". Reference lists of relevant studies, reviews, systematic reviews, consensus papers, and expert guidelines were also searched. Peer-reviewed primary studies, systematic reviews, meta-analyses, technical reviews, consensus documents, and authoritative expert guidelines on CT dose optimization were included. Sources of evidence were prioritized if they covered any issues related to radiation dose metrics, patient size adjustment, acquisition and scan range optimization, reconstruction, artificial intelligence, photon-counting CT, pediatric CT, image quality, or diagnostic performance. Studies not related to CT optimization, highly overlapping with existing literature, or going beyond the scope of this review were excluded.

The search identified 286 records from the electronic databases: PubMed (n = 78), Scopus (n = 96), ScienceDirect (n = 52), and Google Scholar (n = 60). An additional 14 publications were identified through reference-list screening, resulting in 300 records. After removal of 42 duplicates, 258 records underwent title and abstract screening, of which 174 were excluded. Eighty-four full-text articles were assessed, and 14 were subsequently excluded because of insufficient methodological or clinical relevance. A total of 70 publications were included in the final synthesis, as shown in **Figure 1**. Screening was performed by a two-reviewer using database searching, title and abstract screening, and full-text review. Studies were selected based on relevance, methodological quality, clinical applicability, and contributions to the themes predefined in the review.

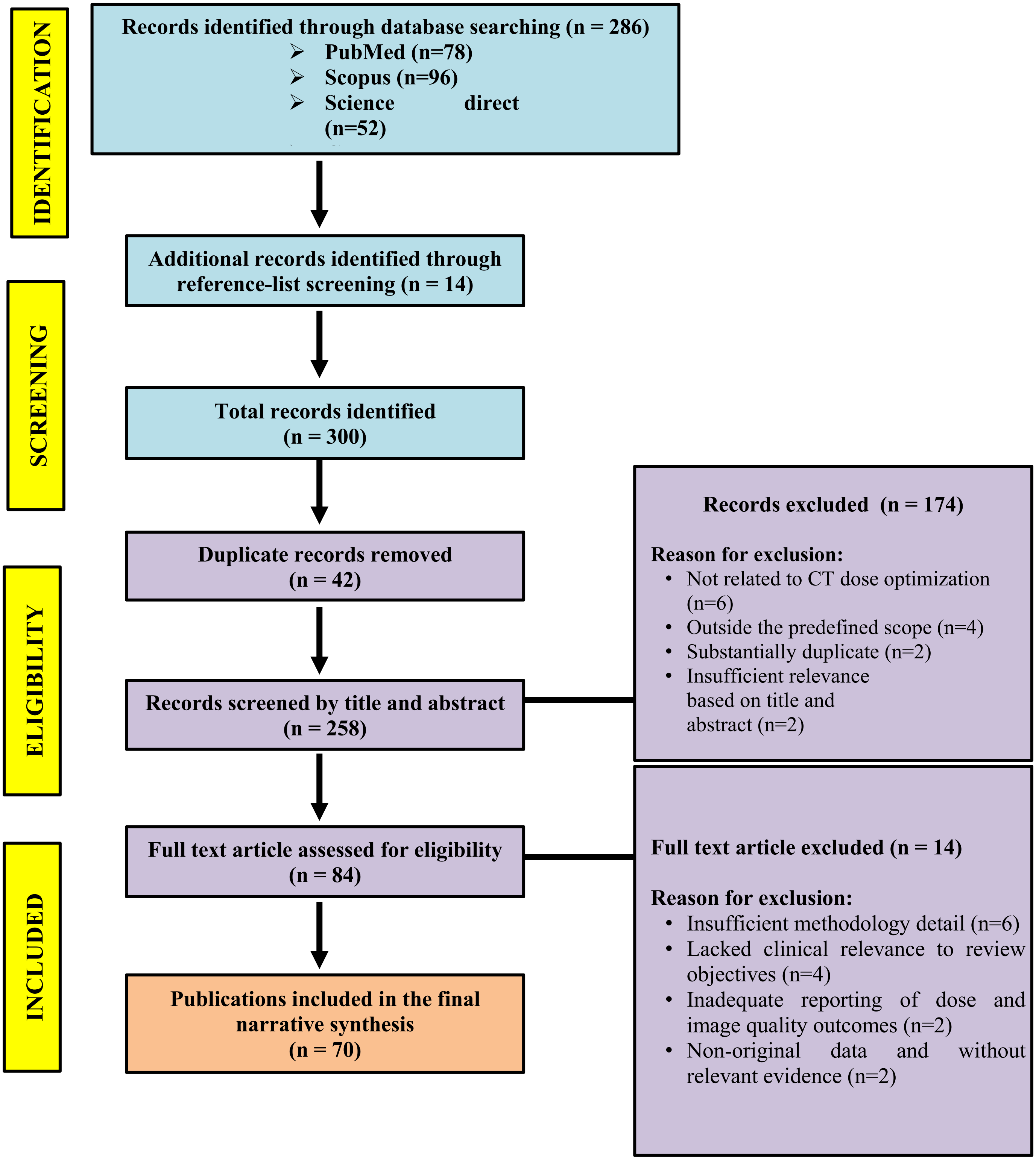

Figure: 1 Flow diagram of the literature identification, screening, eligibility assessment, and inclusion process.

Extraction was done based on a standardized data extraction form, with details of author and year, study design, patient population, type of CT scan, scanner used, acquisition parameters, CT radiation dose metrics, CT optimization approach, image reconstruction technique, image quality criteria, diagnostic performance, key findings, and limitations of studies. The evidence was grouped under eight broad areas: radiation dose metrics for CT scanning and patient-size adjustment, acquisition and CT scanner optimization, iterative reconstruction, artificial intelligence and deep learning reconstruction, photon counting CT, pediatric CT scanning and patient-size optimization, diagnostic image quality and clinical performance and protocol optimization, dosimetry, and quality control. The findings were analyzed qualitatively based on the type of optimization, effect on dose, image quality, clinical relevance, diagnostic significance, limitations, and level of evidence. A meta-analysis was not possible because of the heterogeneity of the studies. The synthesis was conducted based on the difference between simple dose reduction and true dose optimization, where dose reduction is achieved in a manner that maintains image quality and diagnostic value relative to clinical application.

CT RADIATION-DOSE METRICS

CT dose metrics must be interpreted according to what they actually measure. CTDIvol and DLP are standardized scanner-output indices based on reference phantoms [21]. They are not direct measurements of an individual patient's absorbed dose [22]. Patient size, anatomy, scanner geometry, bow-tie filtration, tube potential, tube current, pitch, and scan length all influence the relationship between displayed dose indices and patient dose [22-24]. CT radiation-dose metrics provide standardized measures for evaluating and comparing radiation exposure [21, 22, 25].

The key metrics and their respective uses and limitations are summarized in **Table 1.**

1. **CTDIvol:** For helical CT, CTDIvol is conventionally related to the weighted CTDI (CTDIw) and pitch as follows: CTDIvol = CTDIw / pitch, where CTDIvol is expressed in mGy and describes the scanner output normalized to a standardized reference phantom. It should not be described as the patient dose [21, 22]. For head examinations, the displayed value is commonly based on the 16-cm CTDI phantom; the phantom selection should be documented when comparing protocols [22, 26].

2. **Dose-length product:** DLP is calculated as: DLP = CTDIvol × scan length, where the scan length is expressed in centimeters. DLP is therefore sensitive to both scanner output and the length of the irradiated volume [21]. It is useful for comparing examination-level radiation output, but does not directly account for patient size or organ radiosensitivity [22].

3. **Size-specific dose estimate:** The SSDE was developed to improve the interpretation of CTDIvol by accounting for patient size [23, 26-29]. SSDE = CTDIvol × f size, where f size is a conversion factor determined from an appropriate patient-size metric. AAPM Report 204 established size-based conversion methods for body CT [28], whereas AAPM Report 293 specifically addresses SSDE for head CT [26]. The water-equivalent diameter can provide a more attenuation-sensitive size metric than the simple geometric diameter in appropriate body examinations [29].

4. Organ dose and effective dose: Organ dose is the absorbed dose to a specific organ and is expressed in mGy. It is relevant when evaluating dose to radiosensitive tissues and is generally estimated using anthropomorphic phantoms, Monte Carlo methods, or validated computational models [30, 31]. Effective dose is defined as $E = \Sigma_T w_T H_T$, where w_T is the tissue weighting factor and H_T is the equivalent dose to tissue T [31, 32]. For diagnostic x-rays, the radiation weighting factor is 1, so the numerical value of equivalent dose in a tissue is closely related to absorbed dose in that tissue [31]. Effective dose is a population-based radiation-protection quantity, not a patient-specific prediction of cancer probability [3, 4, 31, 32]. DLP-to-effective-dose conversion coefficients may be useful for broad comparisons, but they depend on the examination region, patient characteristics, scanner and protocol, and the coefficient set used [30, 33]. Consequently, the effective dose should not be presented as an exact individualized patient dose [31].

5. Dose Optimization versus Dose Reduction

Dose reduction and dose optimization are related but not interchangeable. Dose reduction means lowering the radiation output. Dose optimization means selecting exposure conditions that provide adequate diagnostic information for a specified clinical task at the lowest reasonably achievable dose [1, 2].

Aggressive reduction that causes missed lesions, unacceptable noise, loss of low-contrast detectability, or impaired quantitative accuracy is not successful optimization [2, 19, 20, 34].

The optimization process should be approached as a continuous clinical workflow rather than as the application of any single dose-reduction technology. The framework begins with appropriate examination justification, followed by patient-size adaptation and acquisition optimization, dose-efficient reconstruction, objective and subjective image-quality assessment, evaluation of diagnostic performance, and continuous dose monitoring and protocol optimization. This sequence emphasizes that radiation exposure should be reduced only to the extent that diagnostic information remains adequate for the intended clinical task. This process aligns with the ALARA principle while recognizing that image quality is part of patient safety [1, 2, 5, 35].

Metric	What it represents	Unit	Main use	Important limitation
CTDIvol	Average dose index for the scan volume in a standardized CTDI phantom	mGy	Protocol/scanner-output comparison	Not individualized patient absorbed dose; depends on reference phantom
DLP	CTDIvol multiplied by scanned length	mGy·cm	Examination-level scanner-output comparison	Depends strongly on scan length and remains a phantom-based surrogate
SSDE	Size-adjusted estimate derived from CTDIvol using a patient-size conversion factor	mGy	Approximate patient absorbed dose assessment	Still an estimate; must use appropriate size metric/conversion factors
Organ dose	Dose absorbed by a specified organ	mGy	Organ-specific risk assessment	Requires organ-specific measurement; it is not equivalent to effective dose
Effective dose	Weighted sum of organ or tissue equivalent doses for a reference person	mSv	Radiation-protection comparison across procedures	Not an individualized cancer-risk prediction and should not replace organ dose for patient-specific risk

TABLE 1: Radiation Dose Metrics in Computed Tomography

Acquisition and Scanner-based Optimization

Automatic tube-current modulation should be treated as an image-quality control mechanism rather than simply a dose-reduction switch. Correct patient centering is important because AEC systems derive attenuation information from the patient and localizer data [6, 36, 37]. Similarly, reducing tube potential can be advantageous in contrast-enhanced imaging because iodine attenuation increases at lower photon energies [38], but excessive reduction can increase noise and may exceed the tube's ability to maintain adequate output in larger patients [38]. Contemporary CT dose optimization incorporates several acquisition and scanner-based strategies [6, 36, 39]. The strategies, benefits, clinical applications are shown in **Table 2**.

Strategy	Mechanism	Potential benefit	Key limitation / consideration	Clinical application
Automatic tube-current modulation	Adjusts tube current to patient attenuation	Avoids unnecessary output and adapts to body habitus	Requires correct positioning and appropriate AEC settings	Routine adult and pediatric CT
Tube-potential optimization	Uses lower kVp when appropriate, it increases iodine attenuation at lower photon energies	Can reduce dose and improve iodine contrast	May increase noise, it is limited by patient size and tube output	CTA and contrast enhanced CT
Scan-range optimization	Limits acquisition to the required anatomical region	Directly reduces unnecessary irradiated length and DLP	Requires accurate planning	All CT examinations
High-pitch acquisition	Rapid table movement during helical acquisition	Can reduce motion and improve temporal efficiency	Not suitable for every task; image quality and sampling constraints	Cardiac, trauma, pediatric and selected body CT
Spectral filtration/shaping	Removes low-energy photons that contribute little to image formation	Improves beam efficiency	Benefit depends on scanner design and protocol	Routine CT, particularly size- and task-dependent protocols
Organ-based tube-current modulation	Modulates tube current by projection angle/anatomical region	Can reduce peripheral-organ exposure	Requires scanner-specific implementation and validation	Selected head, neck and body examinations

Table: 2. Acquisition and Scanner-based optimization

Iterative Reconstruction

Iterative reconstruction (IR) methods reduce noise by modelling system geometry and/or photon statistics and repeatedly refining the reconstructed image. Hybrid IR is widely used, while model-based IR can provide stronger noise and artifact suppression at the cost of greater computational complexity and, in some implementations, altered image texture. The literature supports meaningful dose reductions with IR in selected examinations [11-15], but dose reduction should not be treated as a universal percentage. The achievable reduction depends on the clinical task, baseline protocol, patient size, reconstruction strength, and image-quality endpoint [11-15, 40].

Artificial Intelligence and Deep Learning Reconstruction

Deep-learning reconstruction (DLR) uses neural networks trained on representative image or projection data to reconstruct or enhance CT images [16, 17]. Depending on the implementation, the algorithm may operate in the projection space, image space, or both. Compared with FBP and some IR methods, DLR can reduce noise while preserving a more natural noise texture and can reconstruct images rapidly. Systematic evidence in head, chest and abdominal CT supports improvements in objective and subjective image quality with potential dose reductions, although reported percentages vary substantially between algorithms and clinical tasks [16, 17]. Therefore, the main advantage of DLR is improved dose efficiency, not a fixed dose-saving percentage, as the scanner can acquire fewer photons while reconstruction compensates for the associated noise penalty [16]. This can be especially valuable when low-dose acquisition would otherwise compromise image quality [16, 41-45]. DLR may also support patient-specific protocol selection and integration with automated quality-control systems [17]. However, DLR has important limitations. Algorithms are vendor- and model-specific; training data may not represent all scanners, patient populations, pathologies, or acquisition conditions [17]. Strong denoising can change noise texture, create an overly smooth appearance, or blur small structures. Complex neural networks are not always fully interpretable, and rare or subtle structures may be altered [17]. The possibility of introducing or suppressing image features is an important theoretical and practical concern [17]. Consequently, objective noise and CNR measurements alone are insufficient; clinical validation should include task-based diagnostic performance, reader confidence, lesion detection, and evaluation of small or low-contrast findings [19, 20].

Current evidence supports DLR as an important optimization technology; however, it should complement not replace, justification, patient-size adaptation, acquisition optimization, and clinical validation [16, 17]. The appropriate dose level remains task-specific [19, 20].

Photon-Counting CT

PCCT uses photon-counting detectors that directly convert individual X-ray photons into electrical signals and classify them according to energy. Compared with energy-integrating detectors, this architecture can reduce electronic noise, permit smaller detector pixels, increase spatial resolution, and provide spectral information from a single acquisition [18]. These characteristics can improve spatial and contrast resolution and increase dose efficiency in selected tasks [18, 46]. Early clinical evidence is seemed to be promising especially in the case of thoracic, cardiac, vascular, musculoskeletal imaging as well as high-resolution imaging [18]. However, PCCT is not automatically a low-dose technology, the amount of radiation dose is based on the diagnostic task, patient size, acquisition mode, reconstruction, and desired spatial resolution [46]. Phantom studies have demonstrated the technological advantages of PCCT, including improved spatial resolution, reduced electronic noise, and enhanced spectral information [18, 46]. Retrospective clinical studies have reported encouraging findings in thoracic, cardiac, vascular, musculoskeletal, and high-resolution imaging applications [18]. Comparative clinical evidence has also evaluated PCCT against conventional CT for image-quality assessment in lung-nodule imaging [47].

Studies have demonstrated the potential for radiation-dose reduction compared with conventional CT protocols while maintaining acceptable or improved image quality in selected clinical applications [46]. These findings indicate that PCCT has potential for improved dose efficiency, particularly when its high spatial resolution, spectral capabilities, and reduced electronic noise are appropriately incorporated into protocol optimization [18, 46]. Current evidence also remains heterogeneous, and broader multicenter studies and protocol standardization are needed [18, 47]. PCCT offers several technological advantages that may improve image quality and dose efficiency. The features, advantages, and limitations are shown in **Table 3.**

Feature	Potential clinical advantage	Current limitation
Direct photon detection	More efficient use of detected photons and reduced electronic noise	Detector physics and count-rate limitations require careful system design
Smaller detector pixels	Higher spatial resolution	Higher spatial resolution can increase quantum noise if not balanced by dose / reconstruction
Intrinsic spectral imaging	Material decomposition and virtual monoenergetic reconstructions	Spectral reconstructions may have task-specific noise and artifact trade-offs
Improved dose efficiency	Potential for lower CTDIvol / SSDE for equivalent diagnostic tasks	Dose savings vary by task, patient size, scanner and protocol
High-resolution imaging	Improved visualization of small/high-contrast structures	Clinical benefit must be demonstrated for the specific diagnostic task
Emerging pediatric applications	Potential dose and image-quality advantages in children	Evidence remains limited and protocols are not yet standardized across vendors

TABLE: 3. Features of Photon-Counting CT (PCCT)

Pediatric CT and Patient-Size Optimization

Children require particular attention because of their greater radiosensitivity and longer expected lifetimes during which radiation-associated effects could manifest [48, 49]. Therefore, pediatric optimization should be based on clinical justification, age/size-appropriate acquisition, automatic exposure control, restricted scan length, appropriate tube potential, and dose-efficient reconstruction [50-54]. A recent systematic review and meta-analysis of pediatric head CT identified protocol modification and iterative reconstruction as important approaches and highlighted substantial heterogeneity according to clinical indications [51]. Patient-size adaptation is preferable to applying a single adult protocol for all children [52].

Weight, age, cross-sectional dimensions, and attenuation-based measures can be used to select or modulate exposure [52, 55, 56]. SSDE is particularly useful because CTDIvol alone does not account for patient size [23].

Routine Organ Shielding

Routine external shielding should not be considered a standard pediatric CT dose-optimization strategy. Contemporary AAPM guidance recommends discontinuing routine gonadal and fetal shielding for X-ray-based diagnostic imaging because shielding can obscure anatomy or interfere with automatic exposure control, potentially compromising image quality or even increasing dose [57, 58]. The AAPM guidance on bismuth shielding advises considering alternative approaches because bismuth can produce artifacts, alter CT numbers, reduce image quality, and interact unpredictably with automatic exposure control [59].

Diagnostic Image Quality and Clinical Performance

A successful optimization strategy must evaluate more than just the radiation output. Image quality is multidimensional and includes image noise, contrast-to-noise ratio (CNR), spatial resolution, low-contrast detectability, artifacts, noise texture, and task-specific diagnostic performance [19, 20]. Subjective image-quality scores should therefore be interpreted alongside objective metrics and, where possible, task-based clinical outcomes [19, 20, 60]. A reduction in image noise or increase in CNR does not prove that a lesion can be detected equally well. Conversely, a modest increase in numerical noise may be acceptable if the relevant diagnostic task remains reliable [20, 21, 60, 61]. Diagnostic image quality should be assessed using both objective and clinical performance measures. The image-quality endpoint and how they are relevant to CT dose optimization are shown in **Table 4**.

Protocol Optimization and Clinical Applications

Protocol optimization should be indication-specific as well. Avoiding unnecessary multiphase acquisitions, limiting the scan range, ensuring accurate patient centering, selecting appropriate tube potential, using AEC, and choosing a reconstruction method appropriate to the task can reduce the dose without sacrificing diagnostic information [1, 6, 11, 36, 38, 62-64].

Endpoint	What it assesses	Relevance to optimization
Image noise	Random variation in CT attenuation values, commonly represented by standard deviation in a uniform ROI	Increases as photon statistics decrease; excessive noise can obscure findings
(Signal to noise Ratio) SNR	Signal relative to noise	Useful for quantitative comparison but does not fully represent clinical detectability
(Contrast to noise Ratio) CNR	Difference between object/background signal relative to noise	More informative than SNR for many lesion-detection tasks
Spatial resolution	Ability to differentiate small high contrast structures	Important for bone, vessels, temporal bone, lung structures and small structures
Low-contrast detectability	Ability to detect subtle attenuation differences	Critical in many abdominal, brain and soft tissue tasks
Noise texture / Noise Power Spectrum	Frequency distribution of image noise	Important because equal numerical noise can appear clinically different
Diagnostic performance	Sensitivity, specificity, lesion detection, reader agreement or task-based performance	Most clinically meaningful endpoint for determining whether dose reduction is acceptable

TABLE 4: Diagnostic Image-Quality and Clinical Performance Endpoints in CT

In contrast-enhanced CT, lower tube potential and spectral reconstruction can improve iodine conspicuity and may reduce the need for higher radiation output or additional phases in selected applications [38, 46, 65, 66]. Dose monitoring should be continuous and ongoing. DRLs can identify unusually high median or typical practice and stimulate protocol review; however, they should be used with clinical indications and patient size information rather than as rigid targets [8, 9, 67]. A process-control approach (measurement, evaluation, feedback, protocol modification, and reassessment) is more consistent with contemporary optimization principles [5,9, 68].

Dose optimization should also consider the clinical indication, since appropriate dose levels differ between routine diagnostic CT and dedicated low-dose applications such as lung cancer screening [64, 69].

Comparative Summary of Contemporary Technologies

AEC and automatic tube-current modulation primarily support patient-specific acquisition by adapting the scanner output to patient attenuation [6]. Tube-potential optimization supports task- and patient-size-specific imaging and can improve iodine contrast while reducing the dose in appropriate patients [38]. Spectral shaping improves photon efficiency through beam filtration, although its benefits remain scanner- and protocol-dependent [24, 46]. Iterative reconstruction uses statistical/model-based refinement to reduce noise and artifacts and can facilitate lower-dose imaging [11-13]. DLR uses neural-network-based reconstruction to improve noise characteristics and dose efficiency. However, its performance depends on vendor implementation, training, reconstruction strength, and diagnostic task [16, 17]. PCCT uses direct photon counting with energy discrimination and provides high spatial resolution, spectral capability, reduced electronic noise, and potential dose savings. Its limitations include cost, availability, technical limitations and heterogeneous clinical evidence [18, 46]. Dose monitoring and DRLs support benchmarking and quality improvement by identifying unusually high or variable practice, but they do not determine the appropriate dose for an individual patient [8]. Contemporary CT dose optimization incorporates multiple acquisition, reconstruction, and monitoring technologies. The summary of contemporary technology mechanisms, optimization roles, benefits, and limitations are in shown **Table 5**.

Critical Synthesis and Evidence Gaps

The literature does not support a single technology as the universal solution to CT radiation exposure [1, 70]. Acquisition strategies reduce the photon burden at the source, while reconstruction strategies recover image quality from lower-photon data [1, 11, 17, 70]. Patient-size adjustment determines whether the scanner output is appropriate for the individual [23, 26, 28, 29], and diagnostic performance determines whether the resulting images remain clinically useful [19].

Technology	Primary mechanism	Typical optimization role	Benefits	Limitations
Automatic Exposure Control / Automatic Tube Current Modulation	Patient attenuation-based tube current modulation	Patient specific acquisition	Reduces unnecessary output, adapts to anatomy	Sensitive to positioning / localizer and vendor specific settings
Tube potential optimization	Lower kVp when appropriate	Task and patient size specific acquisition	Dose reduction, increased iodine contrast	Not suitable for every patient
Spectral shaping	Beam filtration	Improves photon efficiency	Removes less useful low energy photons	Scanner / protocol dependent
Iterative Reconstruct on	Statistical / model-based iterative refinement	Noise and artifact reduction	Enables lower dose acquisition	May alter noise texture, dose saving potential is task specific
Deep Learning Reconstruction	Neural network-based reconstruction	Advanced denoising and dose efficiency	Lower noise, preserved spatial resolution in many studies, rapid reconstruction	Vendor/training dependence, altered texture, possible blurring / diagnostic evidence varies
Photon Counting CT	Direct photon counting with energy discrimination	Dose-efficient high-resolution / spectral CT	High spatial resolution, spectral capability, reduced electronic noise, potential dose savings	Cost, availability, technical limitations and heterogeneous clinical evidence
Dose monitoring / Diagnostic reference levels	Feedback and benchmarking	Quality management process	Identifies outliers and supports protocol improvement	Does not itself determine an appropriate dose for an individual patient

TABLE 5: Comparative Summary of Contemporary CT Dose-Optimization Technologies

Therefore, the strongest optimization strategy is a coordinated combination of justification, acquisition control, reconstruction, and quality assurance [1, 2, 35]. Evidence quality is uneven. Many DLR studies remain single-center, vendor-specific, retrospective, or limited in multicenter clinical validation [17]. Constantly, dose-reduction percentages are difficult to compare directly across studies [16]. DLR evidence is expanding, but prospective task-based validation remains important, particularly for small lesions and low-contrast findings [17]. PCCT evidence is also growing, but multicenter standardization and long-term clinical outcome data remain limited [18, 47]. Future studies should report CTDIvol, DLP and patient-size metrics together; specify the reference phantom and scan length; describe the reconstruction strength; report both objective and task-based image quality; and evaluate diagnostic performance rather than relying solely on noise reduction.

Conclusion

Contemporary CT radiation protection is best understood as optimization of diagnostic information rather than simple numerical dose reduction. CTDIvol and DLP are essential scanner-output metrics but should not be interpreted as individualized patient dose. SSDE provides a more patient-size-aware estimate, whereas organ dose and effective dose address different radiation-protection questions. Modern acquisition techniques, iterative reconstruction, DLR, spectral approaches, and PCCT can improve dose efficiency, but each technology has task-specific limitations. Pediatric CT requires particularly rigorous justification and patient-size adaptation, and routine shielding should not be presented as the primary optimization strategy. In conclusion, the safest CT examination is the one that is justified, appropriately changed to the patient and clinical question, and performed at the minimal dose while preserving the information required for diagnosis.

Declaration of patient consent: Not applicable.

Financial support and sponsorship: Nil.

Conflicts of interest : The authors declare that there are no conflicts of interest regarding the publication of this paper.

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How to cite this article: Sharon R, Jayaiswarya AP, Vayshak KV, Madhesh Kumar M. Contemporary Strategies for Radiation Dose Optimization in Computed Tomography: Balancing Diagnostic Image Quality and Radiation Exposure. 2026;2(3):333–356. doi:10.66159/IJAHS.2026.2309.