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Feasibility of Dose Reduction Using Novel Denoising Techniques for Low kV (80 kV) CT Enterography: Optimization and Validation

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Abstract

Rational and Objectives—To optimize and validate projection space denoising (PSDN) strategies for application to 80 kV computed tomography (CT) data to achieve 50% dose reduction.

Materials and Methods—This retrospective HIPAA-compliant study had IRB approval. We utilized 80 kV image data (mean CTDI_{vol} 7.9 mGy) obtained from dual-source dual-energy CTE exams in 42 patients. For each exam, nine 80 kV image datasets were reconstructed using PSDN (3 levels of intensity) ± image-based denoising and compared to commercial reconstruction kernels. For optimization, qualitative analysis selected optimal denoising strategies, with quantitative analysis measuring image contrast, noise and sharpness (FWHM bowel wall thickness, maximum CT number gradient). For validation, two radiologists examined image quality, comparing low-dose 80 kV optimally denoised images to full dose mixed kV images.

Results—PSDN algorithms generated the best 80 kV image quality (41/42 patients), while the commercial kernels produced the worst (39/42, $p < 0.001$). Overall 80 kV PSDN approaches resulted in higher contrast (mean 332 HU vs. 290 HU), slightly less noise (mean 20 HU vs. 26 HU), but slightly decreased images sharpness (relative bowel wall thickness, 1.069 vs. 1.000) compared to full-dose mixed kV images. Mean image quality scores for full-dose CTE images was 4.9 compared to 4.5 for optimally-denoised half-dose 80 kV CTE images, and 3.1 for non-denoised 80 kV CTE images ($p < 0.001$).

Conclusion—Optimized denoising strategies improve the quality of 80 kV CT enterography images such that CT data obtained at 50% of routine dose levels approaches the image quality of full-dose exams.

Keywords

radiation dose; CT enterography; low-energy CT; image quality; image noise; noise reduction; image denoising; projection-space algorithms; bilateral filtering

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Introduction

Rapid technical innovations in computed tomography (CT) technology have caused a dramatic growth in the annual volume of CT scans, with CT now delivering almost half of the estimated collective dose of radiation exposure in the United States (1). Public health concerns have consequently arisen due to the small theoretical risk of radiation-induced malignancies (2). Despite the controversies over the actual risk of widespread but low-level radiation used in medical imaging, radiologists must adhere to the principle of ALARA, or keeping doses “as low as reasonably possible.”

There are several available methods to reduce CT radiation dose, including the optimized and individualized tailoring of the acquisition parameters (mAs and kVp) to minimize dose and maximize diagnostic information. While both the reduction of mAs and kV values produce a decrease in radiation dose, low kV scanning has additional advantages. Since the signal of iodinated contrast material is higher at lower energies, low kV CT acquisition can also be used as a method to increase disease conspicuity (3,4). In our practice, we have seen dramatic potential for low kV scanning in performing dual-energy (DE) CT enterography (CTE), as iodine signal is increased by a factor of 1.7 at 80 kV (as opposed to 120 kV). Increased iodine signal highlights segmental mural hyperenhancement, which is correlated with active inflammation at small bowel biopsy (5). Since Crohn’s disease is the most common indication for CTE, and because this disease affects young patients that will possibly undergo multiple studies during symptomatic exacerbations throughout their lifetime, radiation dose reduction in this group of patients would be particularly beneficial.

The largest barrier to widespread low kV scanning is the consequent increase in unacceptable image noise (6), which can be reduced not only by increasing tube current or tube voltage, but also utilizing noise-reduction (“denoising”) reconstruction methods (7-12). We recently developed a new projection-space denoising algorithm (PSDA) that takes into account the CT noise model and shows a much better tradeoff between image noise and spatial resolution than the available commercial kernels (13). The purpose of this study is to optimize and evaluate the utility of projection space denoising for CTE exams performed at 80 kV and lower radiation dose. Such an advance would have important implications for lowering cumulative radiation dose in younger patients with chronic diseases and for improving image quality regardless of tube energy.

Methods

Patients and scanning technique

The Institutional Review Board of our institution approved this retrospective study (on 2/12/2008), conducted from data contained in patient databases and archives. Patient inclusion criteria for this study were: (1) signed consent from the patient to use past medical data for research purposes; (2) contrast enhanced dual energy CT enterography examination, conducted for clinical purposes between March 2007 and November 2008.

CT enterography examinations were performed with oral (Volumen; Bracco Diagnostics Inc.) and intravenous (150 mL of iohexol, 300 mg/mL - Omnipaque; Amersham Health, Princeton, NJ - at a rate of 4 mL/sec) contrast material during the enteric phase (beginning 50 seconds after the initiation of contrast material injection) using a dual source CT (DSCT) scanner (SOMATOM Definition, Siemens Medical Systems, Forchheim, Germany) operating in dual-energy mode. Automatic exposure control (AEC) was used for both tubes, with a quality reference mAs of 100 for 140 kV and 425 for 80 kV. The quality reference mAs were selected so that the total CTDI_{vol} of the dual-energy scan was equivalent to our routine single-energy CTE exams. In addition, with this default mAs setting, the CTDI_{vol} of 80 kV is approximately

42% of the total CTDI_{vol} . Using AEC the mAs of the two tubes was modulated independently, and thus the actual total CTDI_{vol} and the distribution between the two tubes varies for different patient sizes. In larger patients, the maximum tube current for the 80 kV tube is reached more often than the maximum tube current for the 140 kV tube. Consequently, the dose from the 80 kV tube is 50% or less of the dose from both tubes. Because this dual-energy CT protocol matches the CTDI_{vol} of our routine single energy CT exams, using only the 80 kV tube from dual-energy CT datasets translates into dose savings of 50% or more, compared to routine clinical imaging using single source CT at 120 kV.

Noise reduction algorithm

Our current approach to projection-space denoising is based upon bilateral filtering (13), which smooths image data using a weighted average in a local neighborhood, with weights determined according to both the spatial proximity and intensity similarity between the center pixel and the neighboring pixels. This filtering is locally adaptive and can preserve important edge information in the image. It is closely related to anisotropic diffusion (14) but is significantly faster. To apply bilateral filtering to CT projection data, the noise characteristics are modeled considering the incident number of photons using AEC data and taking the bow tie filter into account (13). Assuming the detected signal follows Poisson distribution, the sinogram is converted to a dataset representing a map of noise-equivalent number of photons, to which the Anscombe transform (15) is applied to generate a normally-distributed dataset, which is subsequently used for denoising (12,16).

The process for obtaining routine clinical images using commercial reconstruction kernels and for obtaining images with projection and image-based denoising is shown in Figure 1. Raw CT data is denoised offline using a non-specialized personal computer (PC), with denoised data reloaded onto the CT image reconstruction system for image reconstruction using commercially available kernels. Two parameters (σ and ω) were used in bilateral filtering to control the weighting of spatial proximity and intensity similarity and produce different spatial resolution vs. noise tradeoffs [38], with image-based denoising and reconstruction kernels further influencing these tradeoffs.

With the intention of finding the best combination of parameters, the denoising algorithm was applied to the original raw dataset with 3 different σ values (Fig. 1). A single fixed ω value of 5 was chosen because preliminary studies demonstrated visually superior image quality than the other alternate choices. The dataset denoised with the lowest σ value ($\sigma = 1$ and least denoising) was reconstructed using the B40 reconstruction kernel used most often in our routine abdominal CT practice, while $\sigma = 2$ and $\sigma = 3$ datasets were reconstructed using a slightly “sharper” kernel (B45f) to compensate for an anticipated mild amount of image “blurring”. These projection-space denoised datasets were also reconstructed with an commercially-available image-based denoising (3D ORA, Siemens Medical Solutions), as was the original 80 kV non-denoised dataset. 3D ORA stands for “optimized reconstruction algorithm”, and is a non-linear three-dimensional noise reduction filter that does not employ iterative reconstruction using image or projection data. These datasets were compared with an 80 kV original non-denoised dataset reconstructed with the commercial kernels B40f and B25f (a kernel that incorporates a slight amount of 3D image-space denoising; Fig. 1). All images were reconstructed with 1.5 mm slice thickness and 1 mm reconstruction interval.

Image Analysis

This project had two main goals: 1) optimization of the denoising strategies (to investigate which of the several denoising strategies would produce better image quality) and 2) validation of the method (to compare the best denoising strategies with the non-denoised and full-dose

datasets). Accordingly, we divided our experiment in two phases. For the optimization phase, both qualitative and quantitative analyses were performed.

Optimization of denoising strategies - qualitative analysis—The qualitative analysis of the optimization phase was performed by a gastrointestinal radiologist (J.G.F.). By comparing image sharpness, image noise, beam-hardening artifacts and the conspicuity of the bowel wall and enteric inflammation across datasets for each patient, the reader ranked the top four denoising strategies by preference (from 1 to 4) as well as the worst. For each of the top two strategies for each patient, he then rated image quality using a 5-point scale (1 = non-diagnostic due to excessive noise or severe artifacts; 2 = severe artifacts or excessive noise, confidence degraded, diagnosis questionable, 3 = diagnostic, but excessive noise or moderate bowel wall blurriness; 4 = mild noise level or minimal bowel wall blurriness, no change in diagnostic confidence, 5 = minimal noise with “crisp” bowel wall sharpness). Because patient size greatly influences image noise at lower kV utilized in this study, the lateral width of every patient was measured at the level of the iliac crests from the scout images (17,18). Patients were divided in two groups according to their lateral width (≤ 35 cm and > 35 cm), and the best denoising strategy for each of these two groups was determined by selecting the denoising strategy with the most first or second preference rankings.

Optimization of denoising strategies - quantitative analysis—Image noise and image contrast variations across denoising strategies were evaluated by drawing regions of interest (ROI's) with more than 150 pixels. To determine image noise variations, an ROI was drawn in a region of homogeneous fat and the standard deviation (SD) of the mean attenuation value was recorded. To measure image contrast, ROI's were drawn in the renal cortex, with the mean CT number used for comparative analysis. For comparison with full-dose images, we utilized mixed kV images using standard B40 kernel and reconstructed using a 0.7 linear blend of 80 kV/140 kV images, which we employ clinically for CT enterography exams.

In order to quantitatively assess image sharpness, a Matlab[®] tool was implemented to process line profiles traced across the small bowel wall, from lumen to extra-enteric fat. Following manual placement of ROI and line profile tools on a single image, the MatLab[®] tool automatically drew ROI's and line profiles exactly in the same position across all nine denoised datasets. From these profiles we compared 1) the wall thickness based on a full width half maximum (FWHM) calculation (allowing for fractional pixels (5)) and 2) the maximum gradient across the bowel wall, measured as the discrete difference between consecutive pixels' Hounsfield Units (HU) values. These measurements were repeated at three times with slightly different profiles, with the mean value per patient used for analysis, with FWHM measurements normalized to the full-dose image.

Validation of the best denoising strategies—Two different GI radiologists (J.F. and J.H.) performed a separate qualitative analysis of image quality. For each patient, they compared the images rendered by the best denoising strategy for each specific patient and the best denoising strategy for the patient's size group (≤ 35 cm or > 35 cm) with both full dose images and 80 kV non-denoised images (B40 kernel, Figure 2). If these denoising strategies were the same, the second best strategy for the patient was selected for evaluation. They rated image quality for each exam and reconstruction method using the previously described 5-point scale. They were also asked to give a preference between the full dose images (i.e., the mixed kV images) and the half-dose datasets (with ties being allowed), and to rate the diagnostic potential of all four datasets.

The readers independently evaluated randomized CT image datasets at a commercially available workstation in a side by side fashion, panning up and down to compare image quality throughout the imaged volume (Advantage 2.0 Windows; GE Medical Systems, Waukesha,

WI). Readers were blinded to the denoising algorithm and reconstruction kernels utilized to create each dataset. While readers were permitted to change window and level from the default setting (Window 400 Level 40), these changes were made across all datasets being compared.

Results

Patients and Scanning Technique

Of 42 patients, 17 (40.5%) were men (mean age, 46.6; age range, 20-64) and 25 (59.5%) were women (mean age, 46.9; age range, 17-77). Mean patient lateral width at the level of the iliac crests was 35.9 cm, with 20 patients measuring ≤ 35 cm. 14 patients (33.3%) had findings suggestive of active Crohn's disease on full-dose images, while the remaining had normal-appearing small bowel. The mean CT dose index volume (CTDI_{vol}) of CTE examinations was 15.74 mGy. Consequently, the CTDI_{vol} corresponding to the 80 kV tube was ≤ 7.87 mGy, as previously explained.

Optimization of denoising strategies

Qualitative analysis—Projection-space denoising algorithms, alone or combined with the image-based algorithm (3D ORA), generated the best image quality in 41/42 (97.6%) patients. Table 1 displays the number of times each denoising strategy was ranked as first or second by image quality assessment. The denoising strategy that rendered the best images most frequently (12/42, 28.6% of patients) was the projection-space denoising algorithm in its intermediate level of intensity ($\sigma=2$) without superimposed image-space denoising. Each of the six projection space denoising strategies in Table 1 was ranked best at least five times, and the image-space algorithm alone was considered the best denoising strategy in only one patient (2.4%). The original non-denoised dataset reconstructed with the commercial B40f kernel produced the worst images in 37/42 (88.1%) patients.

The best denoising strategy selected for each patient produced diagnostic quality images in all cases, and had a mean image quality score of 4.2 ± 0.8 .

For patients with lateral width ≤ 35 cm, the best denoising strategy was the intermediate level of intensity projection-space algorithm without superimposed image-space denoising ($\sigma = 2$, B45; mean preference rank = 3.2), while for patients with lateral width > 35 cm, the best strategy was the strongest level of projection-space denoising algorithm alone ($\sigma = 3$, B45; mean preference rank = 3.0).

Quantitative analysis—Table 1 shows the relative contrast (in the kidney) and noise resulting from using each denoising strategy, in addition to the bowel wall thickness and sharpness (i.e., gradient across the bowel wall). Contrast of all half-dose datasets was greater than the contrast of the full-dose dataset, because it relied solely on 80 kV data (with 140 kV data lowering the contrast in the full-dose dataset). The greatest reductions in image noise came with projection-space denoising algorithms, which had mean noise levels of 17 – 23 HU, significantly less than the half-dose exam with standard reconstruction kernel (mean noise 33 HU) and similar to the full-dose exams (26 HU, $p \leq 0.003$, except for the dataset denoised with the intermediate intensity level without additional image space denoising; $p=0.112$). There was slight blurring associated with all denoising strategies (greater with projection space methods), resulting in minor increases in relative bowel wall thickness on the order of 1.8 – 9.4% and slight lessening of the maximum CT gradient.

Validation of the best denoising strategies

A summary of image quality assessment by both validation readers is provided in Table 2. The mean image quality for the full-dose datasets was 4.9 ± 0.4 , which was superior to the half-

dose dataset created using the best denoising strategy for each patient (mean image quality rank = 4.1 ± 0.6) and that using the best denoising strategy for the patient's size (mean image quality rank = 4.2 ± 0.7 ; $p < 0.001$ both comparisons), resulting in preference given to full-dose datasets by both readers. However, both denoising strategies using half-dose datasets had significantly superior image quality compared to the non-denoised datasets (mean image quality rank 3.3 ± 0.7 ; $p < 0.001$ for both comparisons).

For both readers, denoised images demonstrated diagnostic quality in all patients. Moreover, the best denoising strategy was considered to provide image quality with no loss of diagnostic confidence (i.e., image quality scores of 4 or 5) in 40/42 (95.2%) and in 39/42 (92.9%) for the readers. In comparison, the low dose 80 kV images without denoising were diagnostic in only 7/42 (17%) and 19/42 (45%), respectively.

Discussion

Our study demonstrates that optimized denoising strategies significantly improve the quality of low kV CT enterography images such that CT data obtained at 50% of routine dose levels is clearly diagnostic and approaches the image quality of full-dose exams. Indeed, using half-dose 80 kV CTE data and the optimal denoising strategy for patient size resulted in no loss of diagnostic confidence when compared to full dose image in 93 - 95 % of patients. In contrast, only 17 - 42% of half-dose 80 kV datasets without denoising resulted in no loss of diagnostic confidence. Our research methodology permitted the paired comparison of full-dose and half-dose 80 kV CT enterography using subjective and quantitative assessment so that the benefits and weaknesses of our approach are transparent. At a time in which the potential risk of radiation induced malignancy is increasingly becoming a matter of public and medical concern, we believe the results of our study are an important contribution to the common effort of the radiology community to optimize the utilization of ionizing radiation. The benefit of these radiation dose reduction techniques is especially significant when applied to CT enterography, as one of the major indications of this exam is inflammatory bowel disease, which frequently affects young people and may demand repeated radiological studies (19).

The novel projection-space denoising algorithm tested in our study proved to be superior not only to the existing commercial kernels, but also to the commercially available nonlinear three-dimensional image-space reconstruction algorithm (3D ORA). The projection-space denoising algorithm, alone or combined with the image-space denoising algorithm, produced the best image quality in 97.6% of cases. These results were corroborated by the quantitative analysis: the greatest reductions in image noise came with projection-space denoising algorithms, which produced mean noise levels of 17 – 23 HU, significantly less than the standard reconstruction kernel (33 HU) and similar to the full-dose exams (26 HU). Moreover, in the vast majority of patients, the extra-time required to post-process the data with the image-space method does not seem to be neither necessary nor significantly beneficial, which adds practicality (and saves time) to the noise-reduction process. Indeed, when the study population was divided in patients with ≤ 35 cm and > 35 cm of lateral width, the optimal denoising strategy for both groups was the projection-space algorithm without superimposed image-space denoising. When this best denoising strategy for the size group was compared with the best strategy to the specific patient (some of which included image space denoising besides projection space denoising), image quality was not significantly different. The projection-space denoising algorithm can usually be performed in 5 minutes using a standard PC. With hardware implementation, the process will be almost in real time. This fast processing time is one advantage of projection space denoising compared to other projection space noise reduction algorithms that involve iterative processes or iterative reconstruction.

The fact that the best denoising strategy for each specific patient did not produce images with quality scores significantly different than the ones produced by the optimal denoising strategy for each size group has important practical value, as it means that the denoising parameters do not need to be optimized for each specific patient. By dividing patients in two size groups (based on their lateral width, which is easily calculated from the topogram of every CT scan), image quality similar to the one obtained by an individual-based selection of parameters can be achieved. Adaptive methods that automatically adjust the denoising parameters according to the attenuation level of the patient are currently under investigation.

Many CT noise reduction methods have been developed, operating on either the raw projection measurement, the log-transformed sinogram, during image reconstruction, or on images after reconstruction (7,10,12,20). In conventional shift-invariant filtration applied during image reconstruction, the suppression of the high-frequency component in the sinogram is performed with a simple assumption that all the measurements are equally reliable, which may result in severe reduction of spatial resolution (7,20). More sophisticated methods have been developed to adaptively smooth the data by taking into account the local statistics in the measurements (7). Some of these methods are currently implemented on clinical scanners, mainly to suppress the streaking artifacts caused by x-ray photon starvation. Many other approaches have also been proposed to incorporate more explicit statistical models and to iteratively restore the log-transformed data by optimizing a penalized weighted least-square or likelihood objective function (12,20). While some denoising or sinogram restoration methods have also been developed to smooth the projection data by taking into account explicit statistical models, either on raw measurement or log-transformed data (10), the proposed approach is non-iterative and practically more feasible. Compared to adaptive filters developed to suppress the streaking artifacts caused by the photon starvation (e.g. Hsieh's adaptive trimmed mean filter (ATM) and Kachelreiss's multi-dimensional adaptive filtering scheme (7,21)), the proposed approach aims to denoise all the projection data while preserving the structural detail.

Most of the published clinical studies on noise reduction filters used reduced tube current to achieve images with less radiation dose (8,9,11). Despite the fact that both the reduction of tube current and tube potential cause a decrease in radiation dose, the reduction of tube potential has an important advantage: the increase in image contrast, which is more pronounced for materials with high effective atomic number, like iodine (4). Several studies have shown that it is possible to reduce tube potential from the usual 120 kV to 80-90 kV and still preserve acceptable image quality (3,22) but, of course, with increased noise. However, to the best of our knowledge, there is no published clinical study investigating the potential of the combination of reduced tube voltage with the utilization of denoising algorithms to reduce radiation dose and increase disease conspicuity in the abdomen simultaneously. The projection-space denoising method evaluated in our study, is also unique, as it is the first in the clinical literature to use bilateral filtering while incorporating the effect of both tube current modulation and bowtie filter effects on image noise.

Our study has some limitations. Only a limited number of kernels (B40f and B45f) were tested with the projection space denoising. Since the radiologists considered the half-dose denoised images to have a slightly inferior image compared to the full dose images largely on the basis of a slight decrease in image sharpness, reconstructing the images with sharper kernels may potentially reduce this problem. The slight blurring of 5% by FWHM measurements of bowel wall thickness (i.e. < 0.2 mm) is clinically insignificant. Since the use of low kV imaging combined with denoising techniques is limited by patient size, we did not test our denoising algorithm in higher kV datasets in a group of larger patients. There is a need for automated tools to predict the lowest possible tube potential and radiation dose for each patient. A general strategy has been proposed (23), but clinical implementation and careful evaluation are still needed. Finally, we did not assess for synergy of projection space denoising with all possible

noise reduction methods (including iterative reconstruction), but we did compare with a commercially-implemented image-based method.

In summary, the novel projection-space denoising algorithm was found to be effective and significantly better than the commercially available kernels and image-space denoising algorithms in reducing CT image noise, so that denoised 80 kV CT enterography images obtained at 50% of routine dose levels become clearly diagnostic and approach the image quality of full-dose exams. The combination of low kV scanning with denoising techniques may have important implications for lowering cumulative dose in younger patients with chronic diseases and for improving image quality.

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Figure 1.

The process for obtaining images of projection space denoising and/or image space denoising with commercial kernels, illustrating the 9 different datasets evaluated. The commercial kernels used to reconstruct images after projection space denoising are listed in parentheses. (3D ORA = three-dimensional non-linear optimized reconstruction algorithm)

Figure 2.

Selected CTE images exemplifying the four kinds of datasets analyzed by the two validation readers. A) 80 kV non-denoised dataset, B) dataset considered the best for this specific patient, C) dataset considered the best for this patient's size group (≤ 35 cm), D) full-dose dataset. Note the significant decrease in image noise from A to B and C. Readers slightly preferred the full-dose, but gave to B) the same image quality score (5). In this patient, there is mild mucosal hyperenhancement in the ascending colon (arrows).

Table 1

Summary of qualitative and quantitative comparison of denoising strategies by contrast, noise and image sharpness (full-width half maximum of ileal wall and maximum image gradient across ileal wall). Values for the full-dose, mixed kV exam are given in grey. Rows above full-dose scan display data for half-dose 80 kV datasets using commercially available kernels (B40 – routine; B25 and 3D ORA – with 3D image-space denoising). Rows below full-dose scan display data for half-dose 80 kV datasets using projection-space denoising algorithms, with or without commercial kernels utilizing image-space denoising. P values refer to the results of paired t-test comparing each denoising strategy compared full dose images.

Denoising Strategy	Qualitative Analysis	Quantitative Analysis			
	# Pts Denoising Strategy in Top 2 by Image Quality	Contrast, renal cortex (p value)	Noise, subcutaneous fat (p value)	Relative FWHM, compared to full-dose (p value)	Maximum Gradient Across Bowel Wall (p value)
Half-dose 80 kV B40	0	340 (<0.0001)	33 (2.38 ⁻⁶)	0.997 (0.800)	134 (<0.0001)
Half-dose 80 kV 3D ORA, B40	1	339 (<0.0001)	28 (0.113)	1.019 (0.107)	128 (<0.0001)
Half-dose 80 kV B25	0	344 (<0.0001)	29 (0.046)	1.018 (0.094)	125 (0.006)
Full-dose Mixed kV, B40	N.A.	290	26	1.000	120
Half-dose 80 kV Denoising (σ=1) with 3D ORA, B40	14	335 (<0.0001)	20 (<0.0001)	1.055 (0.0002)	103 (<0.0001)
Half-dose 80 kV Denoising (σ=1) alone, B40	14	336 (<0.0001)	21 (<0.0001)	1.051 (0.001)	107 (<0.0001)
Half-dose 80 kV Denoising (σ=2) with 3D ORA, B45	12	331 (<0.0001)	19 (<0.0001)	1.063 (0.001)	100 (<0.0001)
Half-dose 80 kV Denoising (σ=2) alone, B45	15	330 (<0.0001)	23 (0.112)	1.056 (0.0004)	102 (<0.0001)
Half-dose 80 kV Denoising (σ=3) with 3D ORA, B45	11	329 (<0.0001)	17 (<0.0001)	1.094 (0.0001)	93 (<0.0001)
Half-dose 80 kV Denoising (σ=3) alone, B45	17	331 (<0.0001)	20 (<0.0001)	1.094 (<0.0001)	96 (<0.0001)

Summary of image preference and image quality scores for full-dose CTE exams and optimally denoised CTE exams by the two validation readers. Image quality was rated using a 5-point scale (1 = non-diagnostic due to excessive noise or severe artifacts; 2 = severe artifacts or excessive noise, confidence degraded, diagnosis questionable, 3 = diagnostic, but excessive noise or moderate bowel wall blurriness; 4 = mild noise level or minimal bowel wall blurriness, no change in diagnostic confidence, 5 = minimal noise with “crisp” bowel wall sharpness; IQ = image quality).

Table 2

	Reader 1 (% of patients)	Reader 2 (% of patients)
Image Quality Assessment		
IQ score, full-dose datasets	4.9	4.8
IQ score, best denoising strategy for each patient	4.1	4.4
IQ score, best denoising strategy for size group	4.0	4.0
IQ score, non-denoised 80 kV dataset	3.1	3.4
IQ score = 5 for full-dose dataset	41 (97.6%)	34 (81%)
IQ score ≥4 for best denoising strategy for the patient	40 (95.2%)	37 (88.1%)
IQ score ≥4 to the best denoising strategy for the patient size group	39 (92.9%)	33 (78.6%)
Same IQ score to the full-dose and best denoising strategy	8 (19%)	26 (62%)
Same IQ score to the best denoising strategies for the patient and size group	38 (90.5%)	28 (66.7%)
Image Preference Comparison		
Full-dose dataset preferred	41 (97.6%)	37 (88.1%)