Monte Carlo evaluation of scatter mitigation strategies in cone-beam CT

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Purpose: To investigate the efficacy of two widely used scatter mitigation methods: antiscatter grids (ASGs) and beam modulating with bowtie filters (BTFs), in combination with subtractive scatter correction or zeroth order normalization phantom calibration, for improving image noise, contrast, contrast-to-noise ratio (CNR), and image uniformity for on-board cone-beam CT (CBCT) imaging systems used for image-guided radiation therapy.

Methods: PTRAN Monte Carlo CBCT x-ray projections of head and pelvic phantoms were calculated for combinations of beam-modulation and scatter rejection methods and images were reconstructed by in-house developed software. In addition, a simple one-dimensional analytic model was developed to predict scatter-to-primary ratio (SPR) and CNR as a function of cylindrical phantom thickness, ASG transmission, and beam modulation with bow-tie filters.

Results: ASGs were found to have slightly negative or no effect on head phantom image CNR and to modestly improve CNR (10%–20%) in pelvic phantom images. However, scatter subtraction and norm-phantom calibration perform better when applied on data acquired with ASGs. Scatter subtraction improves CT number accuracy, but increases noise, and in high SPR/low primary-photon transmission scenarios can dramatically reduce CNR and introduce streaking artifacts. The BTF is found to reduce SPR and image noise, resulting in a better trade-off between CNR and imaging dose, but introduces a circular band artifact.

Conclusions: Our study shows that ASGs have a modest positive impact in pelvic scans and negative in head scans, scatter subtraction improves the HU accuracy but reduces CNR, while BTF has a clearly positive effect. © 2010 American Association of Physicists in Medicine.

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I. INTRODUCTION

Linac-mounted x-ray cone-beam computed tomography (CBCT) imaging systems units have the potential to significantly increase radiotherapy targeting accuracy via daily adaptive interventions to mitigate interfraction setup and tissue-motion errors. However, compared to fan-beam CT (FBCT), CBCT imaging is a relatively new modality with its own signature artifacts, which derive from the large proportion of signal due to scatter, sensitivity to patient motion, and detectors with smaller dynamic range and efficiency.¹ Radiotherapy applications demand highly quantitative imaging performance, including excellent signal uniformity and low contrast structure visualization, so that soft-tissue structures can be accurately segmented by manual or automated techniques.

It is widely observed that the large CBCT scatter signal significantly degrades image quality, through streaking and cupping artifacts, loss of CT number accuracy, and loss of low contrast boundary detectability.^{2,3} To mitigate these effects, several scatter-reduction techniques have been investigated, including antiscatter grids (ASGs),^{4,5} increasing object-to-detector gap,^{6,7} and limiting the field-of-view

(FOV).^{8,9} In addition, algorithmic methods for mathematically subtracting scatter from the sinogram data have been investigated.^{10,11}

ASGs are a well-known, effective means of scatter rejection in projection radiography. Grids always reduce the scatter-to-primary ratio (SPR), thereby improving contrast and image uniformity, but absorb useful primary photons, resulting in increased image noise. ASG impact on CBCT image quality has been found to depend on the anatomic site, SPR, dose, and resolution.^{4,5}

Beam-modulation filters such as bow-tie filters (BTFs) in CBCT offer another method for managing scatter. BTFs are used in radiography and FBCT to reduce the patient dose and to produce a more uniform fluence at the detector, thus reducing the dynamic range requirements. BTFs also improve image quality by better equalizing noise and contrast reduction across the image FOV. In CBCT, BTFs have been found to reduce the SPR by 30%–50% at the projection of the phantom center,^{12,13} while slightly increasing SPR at the phantom periphery. While this SPR reduction mitigates the scatter artifacts in the reconstructed images, the BTF can increase beam-hardening artifacts.

Subtraction of measured or computed scatter profiles from

the projection data restores contrast and CT number accuracy at the expense of increased image noise. Scatter profiles can be measured by variety of techniques.^{8,10,13–16} Experimental techniques have the advantage that they give the scatter magnitude directly in detector signal units, but require the introduction of attenuators, two projections per view in some cases, or require reducing Z-axis coverage. Alternatively, computational methods for estimating scatter corrections can be used, including Monte Carlo (MC) simulation^{5,17} and analytical calculations.¹¹ Both approaches have merits and disadvantages. Analytical calculations may be fast but are limited to simple models of patient shape and composition. MC simulations are more accurate and general (a 6%–10% RMS agreement with measurements reported for a CBCT geometry¹⁷), however, they are computationally demanding, posing an unfavorable trade-off between accuracy and speed. Hybrid deterministic-stochastic techniques^{18,19} show promise of resolving this dilemma.

Selecting efficacious strategies for managing x-ray scatter is essential to maximizing the clinical benefits of CBCT-guided diagnostic and therapeutic interventions. In this investigation, analytical calculations and image reconstructions from synthetic MC data are used to evaluate the impact of subtractive scatter correction (SSC), ASGs, and bowtie beam-modulating filters on image quality. The study end points include contrast, noise, CNR, and image uniformity. Most of the scatter mitigation strategies investigated by this paper have been previously studied, some of them extensively. However, we believe that ours is the first study to systematically assess the cost and benefits of the various combinations of acquisition and postacquisition corrective strategies available to modern CBCT devices. Although this study addresses a specific CBCT system (with a fixed imaging geometry; detector efficiency and resolution; and ASG and BTF geometries), it provides more general insight into interplay of factors that affect the imaging performance of these techniques. In addition, we believe that our study is also the first to quantitatively evaluate the tradeoffs between image noise (and CNR) and image-intensity bias. Finally, the efficacy of the normalization phantom correction approach, which was implemented in a commercial product in 2005,²⁰ has not been previously studied.

II. METHODS AND MATERIALS

II.A. CBCT system description

Our studies were based on the VMS (Varian Medical Systems, Palo Alto, CA) CBCT unit, mounted on the VMS Trilogy medical linear accelerator. The system consists of an x-ray source (VMS G242) and an amorphous silicon flat-panel detector (PaxScan 4030CB, 39.7 cm × 29.8 cm with long dimension parallel to plane of rotation) mounted on robotic arms so that the imaging and treatment-beam central axes are perpendicular. The compact flat-panel detector can acquire images at high frame rate (15–30 frames per second) and has a dual-gain-readout dynamic range of 18 500:1. The system is equipped with a focused 10:1 ASG. A 360° scan with 660 projections is completed in approximately 1 min.

The FOV in full-fan scan mode is a cylinder of 24 cm in diameter and 15 cm in length with the beam limiting collimators fully open. In order to scan larger diameter patients, the transverse FOV is increased by scanning in half-fan mode in which the detector is shifted laterally to acquire only about half of the projection data (at least half is needed to satisfy the data completeness condition) for each acquisition angle. A FOV of 45 cm in the axial plane and 14 cm in the longitudinal direction can thus be achieved. BTFs for either mode can be used. In both full- and half-fan modes, the beam irradiates the full 29.8 cm height of the detector: the 14–15 cm axial FOV is a limit imposed by the vendor's control software. The imaging dose for a typical scan (total mAs of 1260) is 8.3 cGy for full-fan mode, measured on an 18 cm diameter acrylic phantom and 5.4 cGy for half-fan mode, measured on a 30 cm in diameter acrylic phantom.²¹

II.B. One dimensional analytical CNR model

A simple one dimensional (1D) model was developed to predict the influence of SPR on CNR as a function of the diameter of a cylindrical water phantom. The attenuation sinogram $A(x, y, \theta)$ is defined as

$$A(x, y, \theta) = -\ln \left[\frac{P(x, y, \theta) + S(x, y, \theta)}{P_0(x, y, \theta) + S_0(x, y, \theta)} \right], \quad (1)$$

where x, y is location in the detector array, θ denotes gantry angle, P and S denote primary and scatter readings, respectively, exiting the scan subject, while P_0 and S_0 denote the unattenuated primary and scatter readings.

Assuming that the mean energy of primary and secondary photons at the detector to be the same, the relative noise σ/I of a uniformly irradiated x-ray projection as a function of SPR is given by²

$$\left(\frac{\sigma}{I} \right)^2 = \frac{N_p \langle E_p^2 \rangle (1 + \text{SPR}(D)) + N_{\text{add}}^2 \langle E_p \rangle^2}{N_p^2 \langle E_p \rangle^2 (1 + \text{SPR}(D))^2}, \quad (2)$$

where $I = P + S$, N_p and $\langle E_p \rangle$ are the number and the mean energy, respectively, of detected primary photons on the projection of the phantom center, D is the phantom diameter, and $N_{\text{add}} = 5$ photons is the additive Gaussian readout noise corresponding to the reported¹ noise of 3000 electrons for CsI imagers. Assuming that the relative noise σ/I is approximately uniform in the sinogram, the absolute variance $\sigma_{\hat{\mu}}^2$ of the noise propagated into the reconstructed image is given by²²

$$\sigma_{\hat{\mu}}^2 = \frac{\pi^2 \Delta s}{n} \cdot \left(\frac{\sigma}{I} \right)^2 \int_{-\infty}^{\infty} |G(k)|^2 dk, \quad (3)$$

$$G(k) = \begin{cases} k & (|k| \leq k_0) \\ 0 & (|k| > k_0), \end{cases}$$

where $\hat{\mu}$ is reconstructed attenuation coefficient, k is the frequency, $G(k)$ is the frequency domain representation of the filter used in filtered backprojection, Δs is the detector pixel size, k_0 is the maximum frequency, and n is the number of the projections. Equation (3) is valid for both convolution-

based reconstruction in parallel beam geometry and for central transverse-plane in FDK algorithm²³ reconstructions. In this model, the frequency cutoff was chosen to be $k_0 = 0.3N_0$, where $N_0 = (2\Delta s)^{-1}$ is the Nyquist frequency.

For a scan subject of diameter D with attenuation coefficient μ_1 containing a small insert of thickness D_2 and attenuation μ_2 , the contrast reduction (measured insert contrast C relative to the true contrast C_0) is given by^{3,24}

$$\frac{C}{C_0} = \frac{\hat{\mu}_1 - \hat{\mu}_2}{\mu_1 - \mu_2} = 1 + \frac{1}{a \cdot C \cdot D} \ln \left[\frac{1 + \text{SPR}(D) \cdot \exp(-a \cdot C \cdot D)}{1 + \text{SPR}(D)} \right], \quad (4)$$

where $\hat{\mu}_1$ and $\hat{\mu}_2$ are the reconstructed attenuation coefficient values and $a = D_2/D$. For inserts of small size and low contrast, the above equation can be simplified to^{25,26}

$$\frac{C}{C_0} = \frac{1}{1 + \text{SPR}(D)}, \quad (5)$$

showing that the contrast rapidly decreases as a function of phantom size. Combining Eqs. (2) and (5) gives the CNR as a function of phantom diameter. The function $\text{SPR}(D)$ is approximated in this model, by interpolating SPR data vs. cylindrical phantom diameter data derived from our MC code. Based on Eqs. (2)–(4), the effect of SSC, ASGs, and the BTF can be analytically described. The effect of SSC can be included in this model by setting $\text{SPR}=0$ in Eq. (4) and in the denominator of Eq. (2).

ASGs can be included in this model by inserting primary and scatter transmissions ratios, T_{pri} and T_{sc} (ratios of the detected primary and scatter intensities with and without the ASG in place), into Eqs. (2) and (4). Thus, the modified SPR becomes

$$\text{SPR}_{\text{ASG}}(D) = \text{SPR}(D) \cdot T_{\text{sc}}/T_{\text{pri}}, \quad (6)$$

while the number of primary photons detected is $N_{\text{P,ASG}} = N_{\text{P}} \cdot T_{\text{pri}}$. Primary and scatter transmission ratios were considered independent of phantom diameter D . According to this model, an ASG will increase contrast, since it reduces SPR by a factor of $T_{\text{sc}}/T_{\text{pri}}$, while also increasing image noise by reducing the number of detected photons. Based on the above model, the CNR enhancement ratio due to the ASGs, defined by

$$\eta = \frac{\text{CNR}_{\text{ASG}}}{\text{CNR}}, \quad (7)$$

was evaluated as a function of the phantom diameter D for scans with and without the BTF. The BTF influence was modeled by decreasing the SPR by a factor of 0.65, calculated by MC simulation with a 20 cm diameter cylindrical water phantom and a BTF with modulation factor of 6:1 (defined as the ratio of the highest-to-lowest projection intensity with the BTF in place). Primary and scatter ASG transmission ratios were assumed to be unchanged by the inclusion of BTF and phantom diameter.

The number of absorbed primary photons in the detector was calculated by exponential attenuation for a mean primary energy $\langle E_p \rangle = 57.3$ keV, corresponding to a 125 kVp spectrum with 3.1 mm aluminum inherent filtration.²⁷ A detector efficiency of 0.7 was derived from Eq. (9) below and a readout noise² of $N_{\text{add}}=5$ absorbed photons was assumed. The number of emitted source photons was 2.6×10^5 per pixel, which (according to our MC simulations) corresponds to the measured dose of 5.3 cGy at the center of a 16 cm diameter Computed Tomography Dose Index (CTDI) phantom, when no BTF is used.

II.C. Simulation software

The PTRAN MC photon-transport^{28–30} code was used to compute projections of primary and scattered imparted energy distributions in the detector array. The code, initially created for dosimetry applications in brachytherapy, was adapted to the problem of simulating kilovoltage radiographic projections. Phantoms and beam filters are described as set-theoretical combinations of simple analytically describable convex volumes, 3D voxelized arrays, or combinations of the two. The photon transport model includes photoelectric absorption with production of fluorescent photons, Compton scattering (including electron binding effects), coherent scattering, and production of Bremsstrahlung radiation. The calculation of the scattered radiation for CBCT geometry is based on the “bounded next flight estimator,”²⁸ in which every interaction contributes to energy imparted to the detector pixel centered at location (x, y) according to

$$S(x, y) = \sum_{i=1}^{N_h} \frac{1}{\mu(E)} \left(\frac{d\sigma_{\text{incoh}}(E)}{d\Omega} M(E_{\text{incoh}}) + \frac{d\sigma_{\text{coh}}(E)}{d\Omega} M(E_{\text{coh}}) + \frac{d\sigma_{\text{char}}}{d\Omega} M(E_{\text{char}}) \right) \Delta\Omega, \quad (8)$$

where E is energy of the photon entering collision, $M(E) = E \exp(-\int \mu(E) dl) [1 - \exp(-\mu_{\text{CsI}}(E)h/\cos \varphi)]$, $\mu(E)$ and $\mu_{\text{CsI}}(E)$ are the linear attenuation coefficients for phantom material and CsI, $d\sigma_x(E)/d\Omega$ for $x=\text{incoh}$, coh , and char denotes the differential cross section for incoherent scattering, coherent scattering, and characteristic radiation emission following photoelectric absorption, respectively, specified in the medium in which the interaction takes place, l denotes the photon position along its trajectory, $\Delta\Omega$ is the solid angle subtended by the detector pixel, h is the CsI detector thickness, φ is the angle of incidence, and N_h is the number of the simulated photon collisions. The corresponding primary radiation energy impartation distribution $P(x, y)$ is analytically computed, based on ray-tracing calculations through the simulated geometry and line integral evaluations, according to the equation

$$P(x, y) = \sum_{i=1}^{E_0} N_{\text{p}}^0 E_i F(E_i) \exp \left(- \int_l \mu(E_i, l) dl \right) \times [1 - \exp(-\mu_{\text{CsI}}(E_i)h/\cos \varphi)] \Delta E, \quad (9)$$

where N_{p}^0 is the number of source photons per detector pixel

and $F(E_i)$ the source spectrum. This model does not account for photon scattering and emission of characteristic photons in the detector and its supporting structures. National Institute of Standards and Technology (NIST) XCOM-compatible cross sections, atomic form factors, incoherent scattering factors, and mass-energy absorption coefficients were used.³¹

To accelerate the simulation, the scatter projection was computed on a coarse 1 cm grid of detector points and interpolated onto the finer primary-photon grid by bicubic splines. A recent Fourier analysis¹⁶ of typical CBCT scatter distributions demonstrated that nearly all of the spectral power was confined to spatial frequency range below 0.05 line pairs/mm even when spatial square wave modulation was added to the primary x-ray profile incident on the phantom. This corresponds to a Nyquist sampling frequency of about 10 mm.

The prism representation method³² was employed to describe the bow-tie filter. The BTF shape is described by a parallel translation of a polygon (consisting of 700 ordered vertices to describing the BTF cross-sectional shape), which reduces the ray-tracing problem to the simpler problem of finding the intersection of the 2D projection of the ray with the polygon.

A transmission model based on the Day and Dance equation,³³ adapted to the VMS focused grid, was used to simulate the effect of the ASG. This model gives the probability that a photon passes through the ASG without interaction. The probability is a function of photon energy, angle of incidence, grid geometry, and grid material composition. However, our code does not simulate and track scattered and characteristic x rays arising from photon collisions within the grid.

The core PTRAN transport and scoring routines have been extensively benchmarked against experimental brachytherapy benchmark measurements across the 20–1000 keV energy range for a variety of materials and geometries.^{30,34} More recently, we have benchmarked the CT simulator version of PTRAN against extensive beam-stop array scatter-profile measurements using the VMS OBI flat-panel detector. After correction for various competing phenomena and artifacts, e.g., off-axis spectral hardening, veiling glare, beam-block scattering volume perturbations, excellent 3%–5% agreement between measurement and prediction was observed.¹³

II.D. Head and pelvic phantoms

Two mathematical phantoms, representing stylized head and pelvic scan subjects, were designed to evaluate scatter mitigation and correction strategies. The head phantom [Fig. 1(a)] is an elliptical water cylinder (minor diameter of 16 cm, major diameter of 20 cm, and height of 30 cm), containing 2 cm in diameter cylindrical contrast inserts of cortical bone, muscle (6.0% contrast at 56 keV, relative to water), adipose tissue (10.0% contrast), lipid (12.9% contrast), PMMA (10.1% contrast), and breast tissue (6.4% contrast). The pelvic phantom [Fig. 1(b)] is also a elliptical water cylinder (minor diameter of 26 cm, major diameter of 36 cm, and

FIG. 1. Head (a) and pelvic (b) phantoms used in the study. Contrast inserts starting from the lower, going clockwise: Head phantom: 1: cortical bone; 2: PMMA; 3: lipid; 4: adipose tissue; 5: muscle; and 6: breast tissue. Pelvic phantom: 1: prostate; 2: breast; 3: PMMA; 4: muscle; 5: adipose tissue 1; 6: adipose tissue 2; and 7: bone inserts.

height of 65 cm), containing four bony structures and contrast inserts of prostate, muscle, two types of adipose tissue, PMMA, and breast tissue.

660 synthetic primary-photon projections of these phantoms over a 360° scan orbit with 1024 × 768 pixels (0.388 mm pixel size) were computed. Primary-photon energy deposition in each pixel and incident particle fluence was calculated analytically for each photon energy bin, including the attenuating effects of the BTF, ASG, and scan subject, and then integrated over the polyenergetic photon spectrum. A 125 kVp Birch–Marshall spectrum²⁷ with 3.1 mm equivalent aluminum filtration was assumed. Projection data were computed for the following four data acquisition modes: (a) neither ASG nor BTF, (b) ASG but no BTF, (c) BTF but no ASG, and (d) both ASG and BTF. Using MC simulations, the energy-deposition and particle-fluence projections were scaled so as to reproduce the central CTDI doses, measured at our VMS Trilogy CBCT with a CTDI phantom and chamber. For the Varian-recommended techniques (125 kVp, 80 mA, and 25 and 9 ms exposure times per projection with and without BTF, respectively), the number of source photons per pixel was found to be 7.2×10^5 and 2.6×10^5 with and without the bow-tie filter, respectively.

Scatter projections were calculated by Eq. (8) on a grid of 40 points (transverse dimension) × 3 points (axial dimension parallel to the axis of rotation) at the detector surface with a sampling distance of 1 cm. The number of photon histories was 10^6 per projection, giving statistical fluctuations of about 0.5%. To further reduce these fluctuations, the raw scatter distribution was smoothed with a moving average filter nine rows wide. The computation time per projection was 7 s for the analytically calculated primary component (calculated on the full resolution of 2 × 2 binning 1024 × 768 detector grid corresponding to the 40 × 3 scatter grid) and 10 min for the scatter profile. Veiling glare and lag effects of the detector were not taken into account. Compound Poisson-process noise was added to both primary and scatter projections, according to the following equation:

$$P(x, y) = \sum_{i=1}^{E_0} N_s E_i \Delta E, \quad (10)$$

where N_s is a random number selected from the Poisson distribution having a mean given by

$$\lambda_i = N_p^0 F(E_i) \exp\left(-\int \mu(E_i, l) dl\right) \times [1 - \exp(-\mu_{\text{Cst}}(E_i)h/\cos \varphi)], \quad (11)$$

where λ_i is the mean number of photons in the i th energy bin absorbed the detector element.

II.E. Reconstruction software and techniques

An in-house FDK (Ref. 23) code was used to reconstruct the central slice of the image volume. For the head phantom, the reconstruction voxel size was $(1 \times 1 \times 2) \text{ mm}^3$ on a 256×256 grid, while for the pelvic phantom, a 512×512 grid of $(0.8 \times 0.8 \times 2) \text{ mm}^3$ voxels was reconstructed. The maximum spatial frequency of the ramp filter was set to 30% of the Nyquist frequency $(2 \times 0.388 \text{ mm})^{-1}$ for both phantoms.

Four reconstruction protocols were considered: (a) no correction for scatter; (b) reconstruction from primary-photon, scatter-free projections; (c) mathematical SSC; and (d) norm-phantom calibration. Images reconstructed without scatter correction, protocol (a), are included to illustrate the full effect of scatter. The projection images in this reconstruction mode were divided by the air scan to calculate the transmission ratio for every pixel. No beam-hardening correction was applied. In protocol (b), synthetic noise is added to the primary-photon component only. This represents the theoretical best-case scenario in which scatter radiation contributes neither to signal nonlinearity nor to image noise.

Reconstruction with SSC, protocol (c), was performed by subtracting smoothed MC scatter distributions from the full projections. Since noiseless and unbiased scatter distributions (representing the mean value from which compound Poisson random variables are sampled) are subtracted from the noisy projections, SSC yields projections with mean intensities identical with those of protocol (b), but with increased noise (including noise contributions from both the scatter and primary-photon profiles). Because the subtracted scatter profiles are exact (with exception of denoising), our SCC assessment does not include image artifacts and errors introduced by using approximate scatter profile estimates.

Norm-phantom calibration, protocol (d), is used by VMS to calibrate the reconstructed image in the clinically relevant HU range of soft tissues and also as a first-order beam-hardening and scatter correction.²⁰ Each projection of the scan subject is divided by the projection of the cylindrical normalization phantom acquired at the same geometric and technique settings. The reconstruction software operates on the logarithm of this quotient, yielding an attenuation difference image which is then added to the uniform normalization phantom image derived from idealized monoenergetic, scatter-free projections. The full-fan normalization phantom used to reconstruct head-phantom images was a water cylinder of 20 cm in diameter and 35 cm in height. For half-fan reconstructions of the pelvic phantom, a cylinder of 40 cm diameter and 35 cm length was used. In both cases noiseless

calibration-phantom projections were assumed. These specifications approximate the phantoms used in clinical practice with the Varian CBCT.

When BTF projection data are normalized to an air scan (also acquired with BTF), reconstructions exhibit a circular disk artifact (see Figs. 5 and 6) due to the strong beam modulation introduced by the BTF. Residual “ring” artifacts are also visible in reconstructions from primary-photon projections, or projections resulting from scatter subtraction, this time due to varying degrees of beam hardening, induced by BTF. In order to minimize the impact of ring and cupping artifacts on image-quality assessment, projection data were normalized to a 71 keV monoenergetic air scan, instead of the actual polyenergetic air scan. This simple method reduces beam-hardening variations since the additional attenuation that BTF applies to photons passing through the radiologically thin aspects of the scanned object flattens variations in both the projection intensity and average photon energy incident on the detector. For example, our calculations show that the average energy of photons incident on the transverse central axis detector row for a 20 cm diameter water phantom is a relatively constant $71 \pm 4 \text{ keV}$.

For the pelvic phantom an extended FOV was achieved by applying the reconstruction techniques, described above, to synthetic “half-fan” projection data in which the detector was shifted laterally by 15 cm to acquire about half of the projection data over a 360° gantry arc. The half-fan projection data afterwards were rebinned to populate an extended full-fan sinogram representation.³⁵ This method works well for reconstruction of the central-slice images investigated by this study, but rapidly loses accuracy with increasing distance from the central plane. Norm-phantom calibration was also applied for the pelvic phantom in the same way, using simulated half-fan calibration-phantom projections.

II.F. Evaluation of imaging performance

Soft tissue contrast, noise, CNR, and image uniformity were used to assess quality of images reconstructed from the synthetic data sets. Contrast was defined as the difference between the reconstructed values of the contrast insert and water, relative to that of water. To avoid measurement errors due to image noise, contrast measurements were performed on otherwise identical images reconstructed from noiseless data, assuming that the low magnitude of noise does not introduce any bias in the reconstruction process. Image noise was quantified by the standard deviation of image intensity within the 12×12 pixels ROIs located at the center of each contrast insert. To avoid noise measurement error due to systematic image-intensity variation within the ROI, the standard deviation was computed as sum of the squared difference between intensity in the given noisy image and the intensity in the corresponding pixel of the noiseless image. Contrast and noise values were calculated for the adipose tissue inserts, which have reference contrast of 7.1% based on the image formed from primary-photon projection data.

The image nonuniformity index (NUI) was calculated as the difference between the maximum and the minimum reconstructed values of the water voxels, relative to the maximum one,

$$\text{NUI} = 100 \cdot \frac{\hat{\mu}_{98\%} - \hat{\mu}_{2\%}}{\hat{\mu}_{98\%}}. \quad (12)$$

The effective maximum $\hat{\mu}_{98\%}$ is defined as the minimum intensity value that exceeds the intensities of 98% of the water ROI pixels while the effective minimum intensity $\hat{\mu}_{2\%}$ is the maximum intensity exceeding no more than 2% of the water voxels.

Because image-acquisition protocols with and without the BTF deliver different relative three dimensional dose distributions, the noise and CNR values need to be normalized to the same phantom dose. MC simulation was used to calculate 3D dose maps in the head and pelvic phantoms for sinogram acquisitions with and without the BTF. To compute dose maps, the expected-value track-length estimator²⁸ was applied to the $(2.2 \times 2.2 \times 5)$ mm³ head-phantom and $(4 \times 3.5 \times 5)$ mm³ pelvic phantom dose voxels. Assuming that noise is inversely proportional to the square root of dose (despite the difference in x-ray spectra), the normalized image noise $\sigma_{\mu, \text{BTF}}^n$ for BTF data acquisition is given by

$$\sigma_{\mu, \text{BTF}}^n = \sigma_{\mu, \text{BTF}} \sqrt{\frac{D_{\text{BTF}}}{D_{\text{noBTF}}}}, \quad (13)$$

where $\sigma_{\mu, \text{BTF}}$ is the image noise observed using the BTF scanning protocol, while D_{BTF} and D_{noBTF} are the integral dose values for the BTF and no-BTF scan protocols, calculated by summing over the central-slice dose values in the corresponding 3D dose maps.

III. RESULTS

III.A. 1D model for scatter

Figures 2(a) and 2(b) show the CNR enhancement ratio for three hypothetical ASGs for scans with and without BTFs. The grid clinically used by VMS, denoted by ASG2 in Figs. 2(a) and 2(b), has primary and scatter transmissions of 0.75 and 0.31 according to our MC simulations. Our model predicts that ASG2 will have a positive effect on CNR for phantoms with diameters exceeding 18 cm when no BTF is used and 22 cm when a BTF is used. This 4 cm upward shift in the minimum patient diameter for which that ASG improves CNR is observed for the other ASGs as well. For head-size phantoms (diameters between 15 and 20 cm), ASGs result in modest (within 5%) CNR gains when no BTF is used and small losses when a BTF is used [Fig. 2(a)]. In body-size phantoms (diameters larger than 30 cm) full-fan projections filtered by ASGs result in CNR gains of about 10%–15% [Fig. 2(b)].

CNR improvement ratios for ASGs are lower in half-fan mode than in full-fan mode for the same phantom diameter. In half-fan mode, ASGs with primary and scatter transmissions of 0.75 and 0.35, respectively, increase CNR by 8% for 30 cm diameter phantoms (without BTF) and by 3% when

FIG. 2. CNR improvement ratio as function of phantom diameter for scans in full-fan (a) and half-fan (b) scanning modes for three different antiscatter-grid designs (ASG1, ASG2, and ASG3) with and without (w/o) the bow-tie filter (BTF). The transmission ratios (primary, scatter) for each grid design are ASG1 (0.7, 0.28), ASG2 (0.75, 0.31), and ASG3 (0.78, 0.4). (c) CNR vs. phantom diameter with and without subtractive scatter correction compared to CNR achieved in a hypothetical scatter-free imaging system in which only primary photons are detected. (d) Scatter-to-primary ratios for full- and half-fan scanning modes for cylindrical homogeneous water phantoms of varying diameters.

the BTF is used. As with full-fan acquisition, BTFs reduce the effectiveness of ASGs. Finally, our simple model [Fig. 2(c)], based on MC simulated SPR vs. phantom diameter data [Fig. 2(d)] predicts that SSC should leave CNR unchanged due to nearly exact compensation between increased noise [Eq. (2)] and increased contrast [Eq. (4)].

III.B. Scatter and SPR profiles

Figure 3 presents SPR profiles for the central transverse-plane detector row, perpendicular to the CBCT rotational and cylindrical phantom axes, for gantry angles of 0° and 90° . In head-phantom scans with neither BTF nor ASG, the maximum value of SPR at 0° [Fig. 3(a)] is about 1.2 in the shadow of the high density bone insert and 0.65 outside this region. The two almost singular SPR spikes, located near the projected locations of the phantom boundary, are due to the abrupt decrease in primary-photon flux caused by primary photons encountering the lateral aspect of the bone shell. The ASG reduces SPR by a factor of about 2.4 at regions outside the SPR spikes. This reduction is due to the interplay between the 75% primary-radiation transmission and the 31% scattered-photon transmission, characteristic of the VMS grid. Using the BTF without an ASG reduces SPR to 0.43 (SPR reduction factor of 0.57), in the regions adjacent to the bone insert projection, but increases SPR near the phantom periphery. This central axis SPR reduction ratio for the 6:1 VMS BTF is midway between the reported¹² values of 0.86 and 0.6 for 4:1 and 8:1 BTFs, respectively. The combination of the BTF and ASG results in a SPR of 0.28. Slightly lower SPR values are found for head phantom projections at 90° [Fig. 3(b)]. The BTF reduces SPR by 33% at the phantom center and increases it at the phantom periphery. This SPR increase is caused by reduction of the primary radiation component at the periphery due to BTF beam modulation. The reported magnitudes of uncorrected SPR are in reasonable agreement with the literature [0.42 (Ref. 36) to 0.85 (Ref. 5)] as do the ASG reduction factors.³⁷

For pelvic phantom scans [Figs. 3(c) and 3(d)], when neither BTF nor ASG are used, a SPR of 1.28 appears at the projection of the phantom center (4 cm or 102 pixels from the detector end), for 0° projections and nearly 3.5 anterior of the pelvic and femoral head bone inserts for the lateral view (90°). At 0° , the ASG reduces SPR to 0.45 by a factor of 2.84. These values agree with MC calculations of Malusek *et al.*³⁶ (at 0° maximum uncorrected pelvic SPR of 1.51 and SPR reduction factor of 2.84 for a 10:1 ASG). In regions where the primary component is very low, due to the absorption by the bones, larger SPR values were found. Use of the BTF reduces SPR to 0.83 at the center, by a SPR reduction ratio of 0.64. Very high SPR values (SPR of 6.7 when neither the BTF nor the ASG are used) are found for the 90° projection because of very low primary-photon transmission through the bone inserts and lateral separation, which is reduced 2.8-fold by the ASG. The BTF alone reduces the 90° central SPR by 42% (SPR reduction ratio of 0.58), while the combination of BTF and ASG reduces SPR to 1.36. BTF SPR reduction factors for both the 0° and 90° angles agree

with the reported values¹² of 0.78 and 0.54 for 4:1 and 8:1 BTFs, respectively, for a 32 cm diameter cylindrical phantom.

III.C. Dose maps

Figure 4 shows the MC absorbed dose-to-tissue distributions for scans with and without a BTF. 1.3×10^7 photon histories were simulated, resulting in an average statistical uncertainty of 1.8%. The no-BTF head scan (475 mAs) delivers a mean dose of 8.5 cGy and doses of 4.3, 6.9, and 7.3 cGy at the isocenter, lateral skin surface, and anterior skin surface, respectively [Fig. 4(a)]. The dose within the bone shell has a maximum value of 5.1 cGy. The BTF head scan (1320 mAs) delivers a mean dose of 12.7 cGy with doses of 9.1, 9.5, and 7.1 cGy at the three reference points above [Fig. 4(b)]. The mean dose values are used in Eq. (13) to normalize noise in scans with and without the BTF to the same patient dose.

The pelvic scan delivers a mean dose of 4.0 cGy when the BTF is not used (475 mAs) [Fig. 4(c)] versus 6.7 cGy when the BTF is used (1320 mAs) [Fig. 4(d)]. Skin doses of 6.0–6.3 and 6.3–8.4 cGy are found without and with BTF, respectively. When BTF is used, the corresponding skin doses are 8.4 and 6.3 cGy. Doses of 16–18 and 10–12 cGy delivered to the lateral aspects of the femoral heads with and without BTF.

The mean doses are significantly higher than the central or skin-dose values since they are calculated by averaging the dose from all voxels, including the bone structures, where the dose is higher due to bone's larger energy absorption coefficient. However, our MC isocenter dose values for the 18 cm in diameter CTDI acrylic phantom for both BTF and no-BTF cases agree within 2%–7% with reported CTDI measurements²¹ for the same geometric setup and technique settings. Dose values of about 30 cGy were found in the bone shell simulating the skull, higher than that reported by Ding *et al.*³⁸ for the head scan at the similar CBCT unit. However, our calculated value cannot be considered as an estimation of the dose delivered by the scan in patient skull, since the head phantom we use only schematically mimics the human head, both in shape and composition.

The dose values calculated for the pelvic phantom are higher than published ones from OBI scans of prostate patients,³⁹ probably because of the different axial (Z-axis) coverage assumed by the studies (12 cm at isocenter in Ref. 39, versus 20 cm in our simulation).

III.D. Reconstructed images

Figures 5 and 6 present central transverse-plane reconstructed images of the head and pelvic phantom, respectively, while Tables I and II and Fig. 7 present image-quality indexes for the different data-acquisition and reconstruction scenarios. Unmitigated scatter in the head phantom scans (air-scan column of Table I: no BTF, ASG, or correction) reduces contrast (5.4%) by 23% relative to its reference value (7.1%), determined from scatter-free primary-photon projections. This contrast reduction is significantly lower

FIG. 3. Transverse-axis scatter-to-primary ratios for the head and pelvic phantom. (a) Head phantom at 0° projection angle, (b) head phantom at 90° , (c) pelvic phantom at 0° , and (d) pelvic phantom at 90° . The VMS grid (ASG2) was assumed.

than the 55% reduction predicted by Eq. (4), evaluated for the average SPR value (1.2) of source-detector rays intersecting the contrast-insert center. Compared to the primary-photon-only image, the image with unmitigated scatter exhibits slightly less noise (0.95% versus 1.09%) and lower CNR (5.6 versus 6.45). This is because the scattered photons are added to the number of detected photons, thus decreasing the image noise. Unmitigated scatter projections increase the image NUI to 36.2%, compared to NUI of 9.5% for scatter-free imaging. The later value reflects residual cupping due to the beam-hardening effect.

Subtractive scatter correction, as expected, restores contrast and reduces NUI to its scatter-free reference value for all combinations of BTFs and ASGs. However, SSC produces noisier images than either norm phantom calibration or no correction. Depending on the ASG-BTF combination used, SSC image CNRs are 1%–20% lower than the air-calibration (no correction) case and 13%–25% lower than maximum CNR achievable by scatter-free projections. The simple 1D model prediction (identical CNRs for no correction and SSC under conditions of input quantum-limited performance) fails because the model overestimates the impact of scatter on object contrast for asymmetric scan subjects. As shown by Table I and Fig. 7, ASGs improve contrast and image uniformity (6.25% from 5.4% and from 32.6% to 22.6%, respectively) due to the reduction of the SPR. However, the contrast improvement is offset by a corresponding increase in noise, leaving the CNR unchanged or slightly reduced. Combining SSC with the ASG results in a slightly smaller CNR (5.14), essentially identical to the CNR (4.97) achieved by applying SSC to no-ASG data.

When the BTF is used, the minimum NUI achievable falls from 9.5% to 7%, maximum contrast achievable increases from 7.07% to 7.5%, and dose-normalized CNR increases to 7.58 from 6.45, indicating that the BTF mitigates beam-hardening artifacts and image noise as well as scatter-related artifacts. However, as noted earlier in Fig. 5, the BTF gives rise to a ring-shaped artifact which is only partially mitigated by SSC. The BTF is more effective in reducing scatter artifacts than the Varian 10:1 ASG, reducing NUI (15.3%) to less than half that of the no-BTF case (32.6%). The corresponding SPR reductions at the phantom center are 27%–33% (ASG) and 31%–37% (BTF). Compared to ASG alone, dose-normalized BTF CNR is improved (6.9 versus 5.6). In fact, BTF alone without scatter correction produces the highest CNR achievable by any practical combination of mitigation and correction strategies. Applying SSC to BTF projections restores contrast and image uniformity to those of the scatter-free reference image, but at the expense of decreased CNR (5.66 versus 6.9). A properly designed BTF is a highly effective strategy for mitigating scatter that improves all image-quality performance metrics.

Norm-phantom calibration retains the uncorrected image CNR and restores image uniformity (10.6%) to almost that of the scatter-free reference image (9.5%). While it significantly reduces contrast, it reduces noise substantially due to

FIG. 4. Transverse-central plane isodose maps illustrating the absolute dose distribution delivered by the standard Varian Medical System scanning protocols (all using the ASG2 grid) to the head and pelvic phantoms. (a) Head phantom scan without BTF, 475 mAs; (b) head phantom scan with BTF, 1320 mAs; (c) body phantom scan without BTF, 475 mAs; and (d) body phantom scan with BTF, 1320 mAs. Note that the Isodose levels and color codes vary from panel-to-panel. The illustrations were created by the Pinnacle treatment planning system.

adding a noiseless reference-phantom image to the reconstructed patient vs. phantom difference image, leaving the CNR unchanged.

Adding an ASG to the BTF improves contrast and image uniformity and mitigates the ring artifact. However, the ASG results in a CNR loss (5.9 versus 6.9). Applying SSC to data acquired with BTF and ASG slightly reduces CNR (from 5.9 to 5.81) in exchange for maximizing image contrast and uniformity. The SSC-BTF-ASG CNR value is slightly larger than that observed for the SSC-BTF combination (5.66) or SSC alone (4.97). The best image uniformity is given by using the BTF (with or without ASG) in conjunction with scatter subtraction, while the best CNR is given by using BTF without ASG either with no correction or norm-phantom calibration. Overall for head imaging, the best compromise between low NUI and high CNR is no ASG with BTF data acquisition and norm-phantom calibration.

Image quality (Table II and Figs. 6 and 7) for pelvic phantom scans is significantly less favorable than that of head phantom scans. Relative to the scatter-free reference image, unmitigated scatter results in a large cupping artifact (41%) reduces contrast by about 64%, but reduces image noise

(2.05% versus 3.9%) due to the increased number of the detected photons. Overall, scatter reduces CNR (1.24 versus 1.82) by 32%. Similar to the head phantom, SSC restores image contrast and uniformity to reference levels, but results in the highest noise and lowest CNR (23%–30% of the scatter-free reference-image CNR) for all ASG-BTF combinations. In the no-BTF, no-ASG case, photon starvation in lateral x-ray projections is so pronounced that moderately severe noise-induced streaking artifacts are evident. Despite increased noise and artifacts, subjectively, the low contrast inserts are easier to discern in SSC images due to the negative impact of the highly nonuniform image intensity. In contrast to the head phantom and in agreement with our 1D model, the addition of the Varian 10:1 ASG improves contrast (4.9% vs. 2.5%), image uniformity (32.3% vs. 41%), and CNR (1.46 vs. 1.24). Relative to the SSC-only case, applying SSC to ASG projections yields significantly reduced streaking and a moderately improved CNR (0.61 vs. 0.52).

As with the head phantom, addition of the BTF to pelvic scanning improves image quality in all respects (except contrast for which there is slight decrease). The BTF reduces

FIG. 5. Reconstructed images of the head phantom. The gray-scale window is $\pm 20\%$ of the value at the center of phantom. The BTF images assume unscaled doses (unparenthesized CNR and SNR values in Table I).

residual beam-hardening cupping by 40% (NUI of 6.2% vs. 10.9%), and in the presence of scatter, reduces image non-uniformity by 37% (no ASG) to 46% (ASG) and improves CNR (1.6 vs. 1.24 no ASG, and 1.76 vs. 1.46 with ASG). The highest CNR observed (1.76) was with the BTF-ASG combination with no other correction. However, as Fig. 6 reveals, visualization of the contrast inserts is significantly compromised by pronounced residual ring and cupping artifacts. Despite increased noise and reduced CNR, the ASG-BTF-SSC combination, Fig. 6 (last row, second column), has almost no ring artifact and results in the best subjective visualization of low contrast boundaries.

As with the head phantom case, the norm-phantom calibration approach results in the worst contrast, the least image noise, and good-to-excellent CNR (52%–76% of the scatter-free maximum value achievable, see Fig. 7). However, uniformity is improved only for the no-BTF, no-ASG case. In contrast to the head phantom, the norm-phantom calibration does not improve subjective visualization of low contrast structures, illustrating an important limitation of the approach: norm-phantom calibration performance rapidly deteriorates with increasing discrepancy between the calibration phantom (40 cm water cylinder) and scan subject (26 cm by 36 cm ellipsoid) geometries.

Figure 7 graphically illustrates our main results. The top row shows image contrast as a function of image uniformity

(100%–%NUI). For both the head and pelvic phantoms, SSC supports the best image uniformity and contrast, thus yielding the most quantitatively accurate and unbiased distribution of Hounsfield numbers. The addition of the BTF slightly improves uniformity by reducing beam-hardening effects while the ASG adds no benefit. In order of decreasing effectiveness, the strategies for enhancing image-intensity accuracy are SSC+BTF, SSC alone, BTF+ASG without additional corrections, BTF+ASG+norm-phantom, and BTF or ASG without additional corrections. The second row illustrates dependence of CNR (as a fraction of the maximum theoretically achievable value) on image uniformity. For the head phantom, the best compromises between uniformity and CNR is BTF with norm-phantom or no additional correction. Adding an ASG reduces CNR. For pelvic imaging, BTF and ASG together offer the best theoretic compromise between uniformity and CNR with no difference between norm phantom and no correction. Thus, addition of the ASG for large diameter subjects under full-FOV conditions does modestly improve image quality. The graphs clearly illustrate the trade-off that SSC offers between optimal image-intensity accuracy and relatively poor CNR.

IV. DISCUSSION

As the introduction to this paper indicates, scatter is widely accepted as the leading cause of the relatively poor

FIG. 6. Reconstructed images of the pelvic phantom. The gray-scale window is $\pm 20\%$ of the value at the center of phantom. The BTF images assume unscaled doses (unparenthesized CNR and SNR values in Table II).

image quality of CBCT relative to spiral CT imaging. Several groups^{4,5,37} have shown that ASGs can degrade or improve the image quality, depending on the anatomic site and SPR. In agreement with these studies, we find that grids always improve image uniformity and contrast, at the expense, however, of increased noise. In agreement with Kyriakou *et al.*⁵ and Wiegert *et al.*,³⁷ our results show that for small scan subjects the Varian ASG slightly reduces CNR with modest SNR improvement factors for pelvic-size scan subjects.

In general agreement with the literature,^{12,40} the BTF was found to have clearly positive, noncompetitive impact on dose, CNR, and image uniformity, clearly improving the efficiency with which the imaging system utilizes absorbed dose. For the particular designs adopted by Varian, the BTF was more effective in reducing image bias than the ASG, without the CNR loss characteristic of the latter. However, the BTF was found to induce a circular band-shaped image artifact. This artifact is more severe when no scatter correction is applied and is slightly reduced by the ASGs and/or the norm-phantom calibration. The presence of the artifact in images after SSC and in those reconstructed by only primary radiation data shows the need for a beam-hardening correction method specifically designed for the BTF. The simple beam-hardening correction method we used resulted in rela-

tively high NUI (of about 9%–10%), significantly higher than that of a uniform water phantom (2%–3%) (Ref. 41), probably because in the presence of bone structures, the average beam energy on the detector cannot be approximated by a single value. Both our 1D CNR model and MC-based reconstructions show that by reducing SPR, the BTF makes ASGs less efficient. According to the 1D model, the use of the BTF is equivalent, in terms of CNR, to reducing the scan-subject diameter by 3–4 cm.

Our results for mathematical scatter correction are novel and somewhat surprising. Based on our 1D CNR model [Fig. 2(c)], we would have expected SSC to leave CNR unchanged. Our simulations indicate a modest CNR loss for head phantoms and a rather dramatic 2.5-fold CNR loss for pelvic imaging. Reconstructions of the head and pelvic phantom showed that scatter subtraction has a negative effect on the CNR, even when the scatter profile is accurately known, since the improvement in contrast is more than offset by the noise increase. The major cause of the discrepancy is failure of Eq. (4) to predict contrast loss from unmitigated scatter in asymmetric, inhomogeneous phantoms. Essentially, the 1D model significantly underestimates contrast and therefore CNR in the presence of full scatter. Additional MC calculations (not shown here) confirm the accuracy of the 1D model for isocenter CNR in cylindrically symmetric phantoms.

TABLE I. Image-quality indexes of head phantom reconstructed images.

	Image-quality indexes	Air scan calibration	Mathematic scatter subtraction	Norm-phantom calibration	Primary radiation only
Without BT	Contrast (%)	5.4	7.07	4.0	7.07
Without ASGs	Noise (%)	0.95	1.47	0.71	1.09
	CNR	5.6	4.97	5.63	6.45
	NUI(%)	32.6	9.5	11.0	9.5
Without BT	Contrast (%)	6.25	7.07	5.33	7.07
With ASGs	Noise (%)	1.11	1.4	0.95	1.17
	CNR	5.6	5.14	5.6	6.04
	NUI(%)	22.6	9.5	10.5	9.5
With BT	Contrast (%)	5.5	7.5	4.13	7.5
Without ASGs	Noise (%)	0.65 (0.8)	1.08 (1.32)	0.49 (0.6)	0.81 (0.99)
	CNR	8.46 (6.9)	6.94 (5.66)	8.46 (6.88)	9.24 (7.58)
	NUI(%)	15.3	7.1	9.8	7
With BT	Contrast (%)	6.2	7.5	5.3	7.5
With ASGs	Noise (%)	0.86 (1.05)	1.06 (1.29)	0.73 (0.89)	0.92 (1.12)
	CNR	7.18 (5.9)	7.08 (5.81)	7.18 (5.96)	8.13 (6.7)
	NUI(%)	10.6	7	7.5	7
(x.xx) Dose-normalized values					

While many publications confirm the utility of SSC in reducing image bias and cupping, e.g., Bertram *et al.*,⁴² few have looked at its impact on CNR or SNR. One exception is Siewerdsen *et al.*⁸ who showed that empirically derived SSCs reduced cupping with only modest (<10%) CNR losses on body-size phantoms. However, their phantom data reveal only 40% resolution of the cupping artifact and utilize a “full fan” data acquisition technique that truncates the pelvic phantom. Patient images acquired with detector offset (half fan) scanning show subjectively improved image quality, in agreement with our results, but were not quantitatively analyzed. Finally, norm-phantom techniques can substantially reduce scatter-induced image-quality degradation but only when the normalization phantom and scan-subject ge-

ometries closely match, as illustrated by the head phantom centered in the FOV. This strategy is ineffective for markedly asymmetric or inhomogeneous phantoms, e.g., our idealized pelvic phantom, or when the head phantom is positioned off of the FOV center (data not shown). While the norm-phantom reduces cupping and image nonuniformity, it significantly reduces contrast and thus increases image bias.

In summary, our study shows that the BTF and SSC are the most robust and general approaches for mitigating scatter artifacts. The BTF confers a limited benefit but essentially has no downside. Applying scatter subtraction prior to reconstruction can, in theory, completely eliminate nonlinear CBCT response due to scatter. However, contrary to our expectations, it can dramatically reduce CNR in clinical geom-

TABLE II. Image-quality indexes of pelvic phantom reconstructed images.

	Image-quality index	Air scan calibration	Mathematic scatter subtraction	Norm-phantom calibration	Primary radiation only
Without BT	Contrast (%)	2.54	7	1.6	7
Without ASGs	Noise (%)	2.05	13.4	1.33	3.9
	CNR	1.24	0.52	1.2	1.8
	NUI(%)	41	10.9	32.2	10.9
Without BT	Contrast (%)	4.93	7	3.9	7
With ASGs	Noise (%)	3.4	11.4	2.7	4.38
	CNR	1.46	0.61	1.44	1.6
	NUI(%)	32.3	10.8	24.5	10.8
With BT	Contrast (%)	3.3	6.95	2.5	6.95
Without ASGs	Noise (%)	1.59 (2.06)	9.2 (11.95)	1.19 (1.53)	2.3 (3.0)
	CNR	2.07 (1.6)	0.76 (0.58)	2.1 (1.63)	2.97 (2.3)
	NUI(%)	26	6.3	24.3	6.3
With BT	Contrast (%)	5.6	6.95	3.7	6.95
With ASGs	Noise (%)	2.46 (3.19)	7.9 (10.26)	1.6 (2.1)	2.7 (3.5)
	CNR	2.29 (1.76)	0.88 (0.68)	2.26 (1.76)	2.57 (2.0)
	NUI(%)	17.5	6.2	16.6	6.2
(x.xx) Dose-normalized values

FIG. 7. Scattergrams summarizing the impact of various scatter mitigation acquisition and correction techniques on relative image contrast (top row) and contrast-to-noise ratio (bottom row) for the head phantom (left column) and pelvic phantom (right row). The contrast values are normalized to the contrast observed when only primary radiation is detected and CNR is normalized to the value achieved when a bowtie filter but no antiscatter grid is used in conjunction with hypothetical scatter-free projections.

etries, relative to the no-correction scenario, increasing noise to levels where streaking artifacts hinder soft-tissue boundary visualization. Our data support the need to combine subtractive scatter correction with some form of x-ray projection smoothing to compensate for the decrease in SNR following subtraction of the noiseless scatter profile. A number of sinogram denoising techniques, e.g., adaptive low-pass filtering⁴³ and statistically based sinogram-domain restoration^{44,45} techniques, have been investigated, demonstrating that two-fold image SNR enhancement ratios are achievable with minimal loss of spatial resolution.

We conclude this discussion by reviewing the limitations of our study. Our work is limited to one specific CBCT configuration, the Varian Medical Systems on-board imaging system, one set of technique factors, and two different generic phantoms (stylized head and pelvis). However, our stimulated device specifications are applicable to the other

major kilovoltage CBCT system on the market, the Elekta XVI.⁴⁶ Variables that have been shown to impact image quality but were not investigated by us include axial FOV;³ ASG specifications;³⁷ BTF shape;¹² and exposure/projection and reconstruction voxel size.⁴ Regarding exposure/projection, our 1D model reveals that for the exposure conditions assumed by our study, electronic readout noise is small relative to input quantum noise. Hence our results should scale in a straightforward way with patient dose. Second, our results are based solely on simulations, with no direct experimental confirmation. Phenomena that are known to affect image quality but were not incorporated into our study include spatially correlated noise, veiling glare, image lag, and geometric wobble.

In previously presented work, we found that by applying accurate scatter and water-equivalent polynomial beam-hardening corrections eliminates the residual BTF ring arti-

fact in Figs. 5 and 6 in simulations.⁴¹ However, the ring artifact remained when using experimentally acquired sinograms. The residual ring artifact remaining after scatter and beam-hardening compensation was later shown⁴⁷ to be caused by imperfect correction for mechanical wobble. In this simulation study the BTF ring artifact is due solely to inaccuracies in the beam-hardening correction and by the scatter (when scatter was not subtracted). An unexplored opportunity for future research rather than limitation is investigation of adaptive, information preserving sinogram smoothing in the setting of subtractive scatter subtraction. Such a study would require including spatial resolution as a study end point. Prior studies^{4,40} indicate that ASGs and BTFs have little influence on resolution.

V. CONCLUSIONS

Using reconstructions from synthetic data and analytical calculations, we have compared two established approaches for mitigating scattered radiation (bow-tie filter and antiscatter grid) with three corrective strategies (no correction, exact scatter subtraction, and zeroth order normalization phantom calibration) to the unmitigated scatter case for a typical on-board kilovoltage, CsI flat-panel detector cone-beam CT imaging system. Despite its limitations, our study provides a systematic evaluation of clinically practical scatter management strategies for the current generation of CBCT image guidance systems and, more generally, illustrates how realistic MC-based simulations can be used to improve imaging system design while minimizing the number of prototype devices that must be fabricated. In agreement with the literature, our findings show that ASGs only modestly influence low contrast feature detectability, reducing CNR for small scan subjects and improving it slightly for pelvic-size subjects, while improving image uniformity. We have shown that a simple 1D analytic model predicts reasonably well the impact of ASGs on CNR and contrast. The BTF was found to have a clearly positive effect on the dose, image uniformity, and CNR tradeoff, unequivocally improving image quality for all studied end points. Of the corrective strategies, SSC is the most robust and effective strategy for restoring image uniformity and linearity. However, our results demonstrate, contrary to the 1D CNR model predictions, that SSC can dramatically reduce CNR relative to the no-correction scenario, in high SPR scenarios, e.g., extended FOV pelvic imaging. Based on our observations, combining SSC, BTF, and some form of adaptive, feature-preserving sinogram denoising is the most fruitful path for future development efforts.

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