Correction for patient table-induced scattered radiation in cone-beam computed tomography (CBCT)^{a)}

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Purpose: In image-guided radiotherapy, an artifact typically seen in axial slices of x-ray cone-beam computed tomography (CBCT) reconstructions is a dark region or “black hole” situated below the scan isocenter. The authors trace the cause of the artifact to scattered radiation produced by radiotherapy patient tabletops and show it is linked to the use of the offset-detector acquisition mode to enlarge the imaging field-of-view. The authors present a hybrid scatter kernel superposition (SKS) algorithm to correct for scatter from both the object-of-interest and the tabletop.

Methods: Monte Carlo simulations and phantom experiments were first performed to identify the source of the black hole artifact. For correction, a SKS algorithm was developed that uses separate kernels to estimate scatter from the patient tabletop and the object-of-interest. Each projection is divided into two regions, one defined by the shadow cast by the tabletop on the imager and one defined by the unshadowed region. The region not shadowed by the tabletop is processed using the recently developed fast adaptive scatter kernel superposition (fASKS) method which employs asymmetric kernels that best model scatter transport through bodylike objects. The shadowed region is convolved with a combination of slab-derived symmetric SKS kernels and asymmetric fASKS kernels. The composition of the hybrid kernels is projection-angle-dependent. To test the algorithm, pelvis phantom and *in vivo* data were acquired using a CBCT test stand, a Varian Acuity simulator, and a Varian On-Board Imager, all of which have similar geometries and components. Artifact intensities and Hounsfield unit (HU) accuracies in the reconstructions were assessed before and after the correction.

Results: The hybrid kernel algorithm provided effective correction and produced substantially better scatter estimates than the symmetric SKS or asymmetric fASKS methods alone. HU nonuniformities in the reconstructed pelvis phantom were reduced from 220 to 50 HU (i.e., 22%–5%). In the *in vivo* scans, the black hole artifact was reduced by up to 147 HU, a 73% improvement, and anatomical details in the prostate and rectum areas were made considerably more visible.

Conclusions: Radiotherapy tabletops, which are generally flatter and larger than those used for diagnostic CT, can produce significant scatter-related artifacts. The proposed hybrid SKS algorithm accurately estimates scatter from both the object-of-interest and the patient tabletop, and resulting image uniformities and HU accuracies are greatly improved. © 2011 American Association of Physicists in Medicine. [DOI: [10.1118/1.3557468](https://doi.org/10.1118/1.3557468)]

Key words: cone beam computed tomography (CBCT), scatter correction, patient table

I. INTRODUCTION

Kilovoltage x-ray cone-beam computed tomography (CBCT) using amorphous silicon flat-panel detectors provides volumetric imaging capabilities with a single gantry rotation. The modality is well suited for use with linac-based treatment machines and is gaining wide adoption for certain image-guided radiotherapy (IGRT) tasks, including patient positioning and verification. An important application under investi-

gation is adaptive radiotherapy (ART), where treatment plans are adjusted to account for tumor regressions and other anatomic changes that can occur during the course of fractionated therapy. For ART to become clinically viable, however, CBCT images, which generally are not on par with those produced by the diagnostic CT scanners used for treatment planning, must be of sufficiently high quality to allow for organ and tumor recontouring and dose calculation tasks.

A major cause of artifacts in CBCT is the large amount of detected scattered radiation. Scatter-to-primary ratios (SPRs) in the projection data are often much greater than 1, resulting in reconstructed images with soft tissue CT number errors exceeding 300 Hounsfield units (HU).¹ To improve HU accuracies, scatter mitigation hardware, including bowtie filters and antiscatter grids that reduce and reject scattered photons, can be employed.^{2–5} Such hardware is the first line of defense, since it has the advantage of eliminating the Poisson noise associated with the rejected photons. However, a significant amount of scatter often remains, which must be estimated and subtracted from the projection data as part of the reconstruction process in order to produce high-quality images. Most scatter correction methods can be categorized as being based either on estimating scatter using computational models^{6–14} or measuring scatter *in situ*.^{15–20} Methods for measuring scatter *in situ* include collimator shadowing,¹⁵ beam stop arrays,¹⁶ dual scans,^{17,18} and primary beam modulation.^{19,20} While these techniques can be very accurate, they may also have practical limitations such as requiring extra imaging time, additional hardware, reducing the field-of-view (FOV), or being susceptible to patient motion and mechanical instabilities. Hence, the trend remains toward using computational methods.

Among the most practical of the computational techniques is the scatter kernel superposition (SKS).^{6,21} Here, the x-ray cone-beam is modeled as an array of pencil beams that interact with the object-of-interest. For each pencil beam input, a resulting scatter kernel or point-spread function is determined. The total scatter profile, taken as the sum of the scatter profiles from each of the pencil beams, is subtracted from the raw projection data to generate a primary estimate. Unlike other techniques, such as Monte Carlo modeling, the method does not require a first-pass reconstruction, is not susceptible to truncation, and is highly efficient; however, it traditionally has been less accurate. Recently, we introduced a promising SKS enhancement that uses object-dependent adaptive kernels that better model scatter transport through bodylike objects when compared to the traditional slab-based symmetric kernels.^{13,22} After application of the algorithm in a high scatter environment with a maximum SPR > 2, CT number errors in reconstructed images were on the order of $1 \pm 2\%$ (mean $\pm$ standard deviation).

In this work, we show that when the object-of-interest is supported by a radiation therapy patient table, significant CT number errors may still exist after application of the adaptive SKS algorithm. Specifically, a dark region or “black hole” is created beneath the scan isocenter when data are acquired in the offset-detector geometry where the imager is shifted laterally to enlarge the scanned field-of-view. The region above the black hole may be overly bright, in which cases the magnitude of the nonuniformity across the isocenter can be as large as 200 HU (20%). This artifact often coincides with the locations of the prostate and rectum in pelvis scans making the images difficult to use for IGRT applications (Fig. 1).

We demonstrate that the black hole artifact can be traced to scattered radiation produced by radiation therapy tabletops

FIG. 1. Example of the black hole artifact (window=450 HU). The axial slice is reconstructed from a pelvic CBCT data set taken with a Varian Acuity system operating in the offset-detector geometry. The fast adaptive scatter superposition method (fASKS) is used for scatter correction and an analytical model is used for beam-hardening correction (Refs. 13 and 22).

and propose a hybrid SKS method that uses a combination of adaptive asymmetric and slab-based symmetric kernels to correct for scatter from both the object-of-interest and the tabletop. The new method is tested on phantom and *in vivo* data. Image quality is assessed and comparisons are made with gold standard images based on fan-beam acquisitions.

II. METHODS AND MATERIALS

II.A. Cause of the black hole artifact

II.A.1. Patient table hypothesis

Patient tabletops are commonly manufactured from a thin, dense shell surrounding a very low density foam core. The shell is made from a low-atomic-number material having a high fraction of Compton scattering interactions. Compared to tabletops used in diagnostic CT, radiotherapy tabletops are wider in order to be able to center the tumor in the system isocenter and are flatter and thicker in order to provide high stability and less deflection.²³ As illustrated in the lower triangle in Fig. 2, this geometry produces a large amount of detected scatter for views where the tabletop is parallel, or nearly parallel, to the primary x-ray beam. In these cases, the exit surface of the tabletop, which itself a scattering source,

FIG. 2. Two opposing lateral projections shown in the offset-detector geometry. The lower projection, taken at a source angle of 0°, has the tabletop in the FOV. Photons scattered from the exit surface are unobstructed before reaching the detector. The upper lateral projection, taken at a source angle of 180°, does not have the tabletop in its FOV.

FIG. 3. Benchtop system used for the phantom experiments. A pelvis phantom and a patient tabletop segment are scanned together.

is situated in close proximity to the imager leaving an unobstructed path for scattered radiation to reach the detector. For a 360° offset-detector geometry scan, only half of the lateral views are affected. As a result, there exists a close relationship between the projection angle and the amount of tabletop scatter that is created and detected, which we show results in a black hole artifact situated below the scan isocenter.

II.A.2. Phantom scans

To verify the relationship between the black hole artifact and patient table scatter, experiments were conducted on a benchtop system (Fig. 3) having the same geometry as the Varian On-Board Imager (OBI) (Table I). The system used a Varian G242 rotating anode tube operating at 125 kVp and had external filtration consisting of 2 mm of aluminum, 0.5 mm of titanium, and an 8:1 aluminum bowtie filter. The detector was a Varian 4030CB flat-panel imager equipped with a 10:1 antiscatter grid with lamella focused in the axial direction. A pair of tungsten jaws controlled the field size of the x-ray beam, allowing for both cone-beam and fan-beam (or “slit”) imaging geometries. The axis-of-rotation was upright, with a horizontal plate mounted on a driving motor to support a segment of a patient table and/or a phantom. The detector was operated in dynamic gain mode with dark field and gain corrections performed as previously described.²⁴ A 16 cm lateral shift was applied to enlarge the transaxial FOV to 46.6 cm using the offset-detector geometry.

A segment 40 cm in length was cut from of an Exact IGRT radiation therapy tabletop (Varian Medical Systems,

Palo Alto, CA), which was 53 cm wide, 7.5 cm thick, and made with a carbon/epoxy shell of density 1.7 g/cm³ and a polymethacrylimide core of density 0.071 g/cm³. The tabletop segment and a large pelvis phantom (Phantom Laboratories, Salem, NY) measuring approximately 42 cm × 28 cm were scanned together in a cone-beam and a fan-beam geometry. For the cone-beam geometry, the axial extent of the beam was 20 cm at the isocenter, while the axial extent of the beam for the slit geometry was 1.67 cm full-width at half-maximum at the isocenter. To isolate the effect of the patient table on the resulting scatter profiles, the tabletop was dismounted and repeat scans were taken of the phantom alone.

To measure scatter, the slit-scan technique¹⁷ was used. For each projection angle, the difference between the cone-beam and slit (i.e., fan-beam) projection data in their overlap regions was taken. The resulting scatter estimate was smoothed and multiplied by a small calibration factor equal to 1.1, determined using Monte Carlo simulations and experimental measurements, to account for the residual scatter in the slit projections. After extrapolation in the axial direction, the scatter estimate was subtracted from the original cone-beam projection data to generate a primary estimate across the entire projection.

Figure 4(a) shows a lateral projection of the phantom and tabletop taken at a source angle of 8.1° and Fig. 4(b) shows averaged primary and scatter profiles for that projection. The 16 cm shift of the detector results in column 924 mapping to the center of the scan FOV. Note that the primary signals measured with and without the tabletop are similar in the region subtended by the phantom, while the presence of the tabletop significantly increases scatter intensity. As shown in Fig. 4(d), the tabletop nearly doubles the maximum SPR, in this case, to a value above 3.2. The sinogram in Fig. 4(c) maps tabletop-induced SPR as a function of detector column number and projection angle. It is seen that tabletop scatter is most significant for lateral projection angles ranging from −25° to 25°. Reconstructions and corrected data are shown in Sec. III B.

II.A.3. Monte Carlo simulations

To further investigate the relationship between patient table scatter and the black hole artifact, Monte Carlo studies were performed using the GEANT4 toolkit²⁵ (version 4.9.3) with low-energy electromagnetic extension. The simulation setup included a point source and a flat-panel imager with a 0.6 mm thick CsI scintillator. Similar to the experimental setup, the imager was shifted 16 cm laterally for the offset-detector geometry. Neither a grid nor a bowtie filter was simulated. In order to exclude beam-hardening effects, a monoenergetic x-ray beam was used with energy equal to 62 keV, the mean energy of our experimental setup. A total of 240 projections were obtained at 1.5° increments over a 360° rotation.

The first simulation object was a uniform 40 cm × 30 cm elliptical soft tissue (ICRU four-component²⁶) cylinder truncated 5 cm from its bottom so that it could rest on a patient table [Fig. 5(a)]. The idealized tabletop was mod-

TABLE I. Varian OBI geometry.

Parameter	Value
Source-to-isocenter distance	100 cm
Source-to-detector distance	150 cm
Source-to-bowtie distance	15 cm
Maximum transaxial FOV	46.6 cm
Detector matrix size	1024 × 768 (with 2 × 2 binning)
Detector pixel size	0.388 mm (with 2 × 2 binning)

FIG. 4. Scatter and primary measurements from the pelvis phantom scanned with and without the tabletop. (a) Raw projection data taken at a lateral source angle = 8.1° . In the offset geometry, the tabletop shadows the left portion of the projection. Column 924 corresponds to the center of the FOV. A small air gap between the tabletop and the phantom in the upper part of the projection is visible. (b) Measured scatter and primary profiles averaged over the 46 central rows of the detector for the same lateral projection. The patient tabletop significantly increases the total amount of scatter while the primary signal in the region subtended by the phantom remains unchanged. The vertical scale is in analog-to-digital readout units of the detector. (c) Sinogram showing tabletop-induced SPR as a function of projection angle and detector column. The intensity of each sinogram row indicates the additional SPR, averaged over the 46 central rows of the detector, due to the tabletop for a given projection angle. (d) SPR profiles with and without the table for the projection angle = 8.1° . The tabletop nearly doubles the maximum SPR.

eled as a 5 cm thick, 50 cm wide polycarbonate slab with a uniform density of 0.38 g/cm^3 , which equals the average density of typical radiation therapy tabletops. All scattering events were tagged according to where the interaction occurred, thus allowing for each energy deposition event into the CsI detector to be categorized as resulting from either primary transmission, object-only scatter, tabletop-only scatter, or combined object and tabletop scatter. Thus, four projection images were generated for each projection angle. To investigate the effects of bony structures on scatter distributions and on cupping artifacts, the object was modified to include cortical bones²⁶ resembling the hips and vertebral disks of a lower pelvis [Fig. 5(b)] and the entire simulation was repeated. For both simulations, tabletop and object scatter profiles and reconstructed images were analyzed.

II.B. Hybrid kernels for object and table scatter correction

The SKS algorithm was modified to correct for both object and tabletop scatter. Traditionally, the algorithm has used symmetric kernels whose amplitudes and shapes are determined by the scatter point-spread functions produced by directing a pencil beam through uniform slabs of material of varying thicknesses. The estimated scatter is computed by convolving primary estimates with the kernels²¹

$$I_s(x, y) = I_p(x, y) ** h_s(x, y), \quad (1)$$

where I_p and I_s are scatter and primary estimates, respectively, and h_s is the scatter kernel. Because neither I_s nor I_p is known *a priori*, both must be simultaneously solved for in an

FIG. 5. (a) Cross sections of the tabletop and the soft tissue phantom used in the simulation studies. The elliptical cylinder measures 40 cm $\times$ 30 cm in the transaxial direction and is truncated 5 cm from its bottom. The tabletop is a 50 cm wide and 5 cm thick. (b) The elliptical phantom with bony structures added. Note that the jagged edges in the structures are display artifacts caused by the rendering program (VRMLVIEW) used to visualize the GEANT4 geometries.

iterative fashion.^{8,27} Convergence usually occurs after three to four iterations.²²

Recently, we have shown that the symmetric kernels derived from uniform slab responses often do not accurately model scatter transport through bodylike objects that tend to be more elliptical in shape. Generally, scatter is overestimated toward the center of the object and underestimated toward its edges, leading to doming artifacts and other HU inaccuracies in the final reconstructions.^{13,22} The cause of the problem is illustrated in Fig. 6 by the simple example of a primary ray originating between the center and left edge of an elliptical object. Due to exit side curvature, scattered photons that are produced by the primary ray are more likely to be absorbed toward the center of the projection where the object is thicker than toward the edges. However, under the slab approximation, the object is modeled as having a uniform thickness which leads to an overestimate of scatter toward the center of the projection and an underestimate toward the left edge as shown.

To address this weakness, we have proposed the adaptive scatter kernel superposition (ASKS) algorithm using asymmetric kernels whose shapes and amplitudes are modulated according to local thickness changes in the vicinity of the primary pencil beam.^{13,22} The resulting estimate was shown to be more accurate than that produced by the symmetric kernels, and it can be computed efficiently by expressing the object-dependent adaptive kernels h_a in terms of the traditional symmetric kernels h_s

$$h_a(x-x', y-y') = h_s(x-x', y-y') \cdot (1 - \gamma \cdot (\tau(x, y) - \tau(x', y'))), \quad (2)$$

where τ is the water-equivalent thickness profile of the object, derived from the primary estimate. γ is a proportionality coefficient whose value depends on the imaging geometry and other empirical considerations. The above kernels are quasistationary and allow for the convolution to be performed rapidly in the Fourier domain. The method is denoted as fast adaptive scatter kernel superposition (fASKS).

The fASKS algorithm was shown to be effective for bodylike objects. However, in subsequent studies, we have found that the algorithm underestimates scatter from the patient table for lateral projections. This is because scattered

FIG. 6. The scatter kernel produced by a primary pencil beam is modulated by local thickness variations in the object. Using Monte Carlo simulations, a pencil beam is directed between the left edge and the center of an elliptical object. The resulting scatter response (measured) is asymmetric. This response is better modeled by the adaptive kernels (ASKS) than by the symmetric kernels (SKS).

photons that exit the tabletop are unobstructed before they reach the detector, thus violating a premise of the adaptive model that these photons should be attenuated in part by the object. As shown below, due to its extended rectangular geometry, the tabletop behaves more like a discrete slab and the symmetric kernels can better represent its scatter properties.

To model both object scatter and tabletop scatter, we propose a hybrid method that combines use of symmetric and adaptive kernels. Figure 7 illustrates the rationale for the algorithm. As shown in Fig. 7(a), the fASKS algorithm fails for a lateral primary ray directed through the tabletop because the entire tabletop mass is assumed to be concentrated beneath the center of the object when, in fact, the main scattering sites are located at the entrance and exit surfaces of the tabletop [Fig. 7(b)]. The response from the entrance surface is asymmetric and can be modeled by the fASKS algorithm, while scatter that emerges from the exit surface is unblocked and has a symmetric response. Consequently, the total scatter response for this lateral ray is best modeled using a mixture of symmetric and asymmetric kernels with weighting favoring the symmetric kernels since the exit surface is closer to the detector. As the angle of the ray-of-interest increases, weighting starts to favor the adaptive kernels. This is partly because tabletop-induced scatter becomes less significant relative to object scatter due to primary ray attenuation [Figs.

FIG. 7. Rationale for the hybrid kernel algorithm to correct for tabletop scatter. (a) For a lateral ray directed horizontally through the tabletop, the adaptive algorithm (fASKS) incorrectly assumes that all tabletop mass is concentrated beneath the object, thus yielding an asymmetric scatter response. (b) In actuality, most rays are scattered either at the entrance or exit surface of the tabletop. The scattered rays emanating from the exit surface are unblocked yielding symmetric responses, while scattered rays emanating from the entrance surface have asymmetric responses. [(c) and (d)] As the primary ray angle increases, the tabletop scatter becomes less significant relative to object scatter and the overall response is better modeled by the fASKS algorithm.

7(c) and 7(d)] and partly because tabletop scatter can itself have an asymmetric response that is modeled by the fASKS algorithm [Fig. 7(d)].

Given the above rationale, the algorithm can be described as follows: For primary rays that *only* pass through the object, the fASKS adaptive kernels h_a , described by Eq. (2), are used. However, for primary rays that pass through only the tabletop or pass through both the tabletop and object, a hybrid scatter kernel h_{hybrid} , comprising both the adaptive and symmetric kernels, is used

$$h_{\text{hybrid}} = w \cdot h_s + (1 - w) \cdot h_a. \quad (3)$$

The weighting coefficient w ranges between 0 and 1 and depends on the tabletop geometry and gantry angle. As suggested by Fig. 7, the general rule of thumb is that w has higher values for lateral rays that pass through the tabletop and lower values for the more vertical rays whose scatter responses are adequately modeled by fASKS alone.

The algorithm preserves the amplitude of the original scatter kernels at their origins $x=x'$, $y=y'$ [Eq. (2)], since, as shown in Fig. 6, both the symmetric and adaptive kernels have the same amplitude at the location of the primary pencil beam. Therefore, as with the design of the adaptive kernels

themselves, the effect of the hybrid kernel mix is to change the kernel shape in locations away from the primary pencil beam.

II.B.1. Weighting function

The optimization of weighting function parameters is based on measurements taken with calibration phantoms. As illustrated in Fig. 7, the symmetric component is peaked near the gantry angle where the x-ray beam is laterally directed through the tabletop and then declines fairly rapidly as the gantry moves away from that angle. Based on this behavior, we model the weighting function w as a Gaussian

$$w(\theta) = A \exp\left(-\frac{(\theta - \theta_0)^2}{2\sigma^2}\right), \quad (4)$$

where θ is the projection angle and θ_0 is the angular location of the maximum symmetric component.

To determine the weighting function parameters, a primary error sinogram of the calibration phantom is first calculated by subtracting the primary estimate made with the adaptive (fASKS) kernels alone from the ground-truth primary estimate made using the slit-scan technique. The weighting parameters are then optimized to minimize subse-

FIG. 8. Hybrid SKS correction algorithm. Each projection is divided into two regions: One shadowed by the tabletop (R_1) and one not shadowed (R_2). These two regions are convolved with the hybrid and adaptive kernels, respectively. The two scatter estimates are then summed to form a complete scatter estimate.

quent primary errors after application of the hybrid algorithm. The function is evaluated in the range of $[\theta_0 - 180^\circ, \theta_0 + 180^\circ]$ to ensure continuity upon angular wrap-around. We have found that the same parameters for w can be used for different object sizes, but that w may change for different tabletop geometries.

Using the pelvis phantom for calibration, the following weighting function parameters were determined for the Exact IGRT tabletop: $\sigma = 25.5^\circ$, $A = 0.8$, and $\theta_0 = 7^\circ$. The *in vivo* data were acquired using a table from a third-party manufacturer and the parameters were recalibrated for that tabletop. Note that since the hybrid kernels are only applied to pencil beams located within the part of each projection that is shadowed by the tabletop, it is important to efficiently identify and segment the tabletop. We use *a priori* knowledge of the tabletop shape, along with knowledge of the gantry angle and the tabletop position as reported by the linac, to map the shadowed region for each projection.

II.B.2. The complete hybrid kernel algorithm

For each projection image, the region not shadowed by the tabletop is convolved with the fASKS kernels while the shadowed region is convolved with the hybrid kernels. The complete algorithm is illustrated in Fig. 8 and is outlined as follows.

For each projection,

- (1) Generate a weighting factor $w(\theta)$ where θ is the projection angle.
- (2) Project the patient tabletop onto the detector using *a priori* knowledge of the tabletop position and shape. Divide the detector area into two regions: One shadowed by the tabletop (R_1) and one not shadowed (R_2).
- (3) Iteratively estimate scatter and update the primary signal.
  - (a) Initialize the primary estimate across the entire projection.
  - (b) For pencil beams in region R_1 , convolve the primary estimate with the hybrid kernels (h_{hybrid}) according to Eq. (3).

- (c) For pencil beams in region R_2 , convolve the primary estimate with the adaptive kernels (h_a) using fASKS.
- (d) Sum the scatter estimates generated from the R_1 and R_2 primary beams to produce the complete scatter estimate.
- (e) Subtract the scatter estimate from the raw data signal to generate a new primary estimate.
- (f) Loop back to step (b) until convergence.

To increase computational speed, the original projection data can be downsampled to “superpixels” since the scatter signal is largely composed of low frequency components. For our application, the superpixel size was 10 mm in the axial direction and 4 mm in the transaxial direction, corresponding to factors of 26 and 10 reductions in matrix size, respectively. After the scatter profile is computed, it is upsampled and subtracted from the original projection data at full resolution.

The hybrid kernel implementation [steps (b)–(d)] requires two convolutions, one with the hybrid kernels and one with the adaptive kernels

$$I_s = (T \cdot I_p) * h_{\text{hybrid}} + ((1 - T) \cdot I_p) * h_a, \quad (5)$$

where T is a binary mask that has the same dimensions as the downsampled projection data and indicates the presence of table-shadowed superpixels. The mask is defined such that $T=1$ for table-shadowed superpixels and $T=0$ for unshadowed superpixels.

Substituting for h_{hybrid} using Eq. (3), the convolution is written in terms of h_a and h_s

$$\begin{aligned} I_s &= w \cdot (T \cdot I_p) * h_s + ((1 - w) \cdot (T \cdot I_p) \\ &\quad + (1 - T) \cdot I_p) * h_a \\ &= (wT \cdot I_p) * h_s + ((1 - wT) \cdot I_p) * h_a. \end{aligned} \quad (6)$$

We can further expand the equation by writing the adaptive fASKS kernels (h_a) in terms of the symmetric kernels (h_s) using Eq. (2) and rearranging so that only two convolutions with stationary kernels are required

FIG. 9. (a) Sinogram showing the increased SPRs caused by the tabletop for the soft tissue elliptical phantom simulation. Column 924 corresponds to the center of the scanned FOV. (b) Decomposed SPR curves for projection angle = 0° demonstrate that tabletop scatter is more than twice that of object scatter for this lateral projection. (c) Tabletop SPR sinogram for the bony simulation. (d) Decomposed SPRs for the bony simulation for projection angle = 0°. The presence of the bones increases the maximum SPR by more than two times. Note that object+tabletop scatter refers to multiple order scattering events occurring in both the object and tabletop, while tabletop-only or object-only scatter refers to single and multiple order events occurring in either just the tabletop or object.

$$\begin{aligned}
 I_s &= (wT \cdot I_p) * h_s + (1 - \gamma\tau)((1 - wT) \cdot I_p) * h_s \\
 &\quad + \gamma(\tau(1 - wT)I_p) * h_s \\
 &= ((\gamma\tau + (1 - \gamma\tau)wT) \cdot I_p) * h_s \\
 &\quad + (1 - \gamma\tau)((1 - wT) \cdot I_p) * h_s,
 \end{aligned} \tag{7}$$

where γ , τ , w , τ , I_p , and h_s are as defined previously. As with fASKS, these convolutions can be performed efficiently in Fourier space. Thus, the algorithm is as computationally efficient as fASKS.

The complete FDK reconstruction process starts with detector gain and offset corrections²⁴ followed by the hybrid scatter correction described by Eq. (7). The scatter-corrected data are then log-normalized using an air scan to create water-equivalent thicknesses for each projection element. Beam-hardening correction is then applied using an analytical algorithm based on models of the beam spectrum, external filtration, bowtie filter, and detector response.¹³ This is

followed by FDK convolution and backprojection²⁸ and scaling to Hounsfield units. Ring correction is applied to the final images.²⁹ For our experiments, a transaxial FOV with a diameter of 466 mm was reconstructed on a 512×512 matrix. The reconstructed slice thickness was 2.5 mm.

II.C. Scatter correction evaluation

II.C.1. Phantom studies

The algorithm was first tested on the pelvis phantom that was scanned using the benchtop system. Data were acquired with and without the patient tabletop and corrected with the symmetric (SKS), the adaptive (fASKS), and the hybrid kernel algorithms for comparison. Scatter estimation accuracy was evaluated by calculating the primary estimation error as follows:

$$\text{Err}_{\text{primary}} = \frac{I_p - I_{p0}}{I_{p0}} \times 100\%, \quad (8)$$

where I_{p0} is the gold standard primary signal given by applying the slit-scan scatter correction to data taken without the table. Artifact intensities and HU accuracies were assessed in the reconstructed images.

II.C.2. In vivo patient data

In vivo pelvis scans were acquired in the offset-detector geometry using a Varian Acuity simulator and a Varian OBI system, both of which were equipped with commercial patient tables. The OBI and Acuity CBCT systems have comparable scan geometries and use the same x-ray source (Varian G242 tube) and flat-panel detector (Varian 4030CB imager). The main difference between the two systems is that the OBI has 2.7 mm total aluminum beam filtration along with an 8:1 aluminum bowtie filter, while the Acuity has 7.2 mm total aluminum filtration along with an 8:1 plastic bowtie filter. The data acquisition parameters for the Acuity scans were 125 kVp, 2 mA s per projection, and 679 projections. For the OBI scans, the acquisition parameters were 125 kVp, 2 mA s per projection, and 626 projections. Acuity scatter measurements and reference images were obtained using the slit-scan technique. Slit scans were not acquired on the OBI system, so the OBI reconstructions were compared to diagnostic quality images acquired using a state-of-the-art helical CT system (General Electric, Waukesha, WI). *In vivo* images were evaluated for CT number accuracy, uniformity, and anatomical structure visibility.

III. RESULTS

III.A. Simulations

The purpose of the simulation studies was to investigate the link between patient table scatter and the black hole artifact. The sinograms in Fig. 9 show tabletop-induced scatter as a function of projection angle for the uniform elliptical phantom without [Fig. 9(a)] and with [Fig. 9(c)] the bones added. For both cases, the high SPR region is concentrated about the 0° projection angle. The addition of the bony structures did not greatly affect the magnitude of the scattered signal (data not shown), but the SPR increased due to the higher attenuation of primary rays by the bones. Figures 9(b) and 9(d) plot the SPRs resulting from photons scattered by only the tabletop (*tabletop-only*), by only the object (*object-only*), or by both (*object+tabletop*) for the highly impacted 0° view. For the water-only phantom, tabletop scatter was close to six times higher than the primary signal and two to three times higher than object scatter. The addition of the bony structures increased the maximum tabletop SPR to over 15.

In part, due to the lack of a bowtie filter and antiscatter grid, the simulated SPRs were higher than those measured experimentally. Nevertheless, all results show the important and unique characteristic that tabletop scatter is most significant for lateral or nearly lateral projection angles.

Reconstructed images from the boneless simulation are shown in Fig. 10. The image from the primary signal alone is, as expected, uniform except for artifacts resulting from statistical noise. The reconstruction of the primary plus object-only scatter signals shows scatter-related cupping artifacts with a central doming pattern typical of offset-detector scans.³⁰ In contrast, the image reconstructed from primary photons plus tabletop-scattered photons demonstrates a distinct black hole artifact below the isocenter. The vertical bright-to-dark transition across the isocenter has a nonuniformity exceeding 700 HU or 70%.

Resulting images from the bony simulation are shown in Fig. 11. Similar artifact patterns to those from the water-only simulation are demonstrated but with increased intensities. The black hole artifact is characterized by a bright region measuring over 400 HU and a dark region intensity of −600 HU. Overall, the bright-to-dark transition is 400 HU higher than that for the water-only simulation. We speculate that this 67% increase is caused by the higher SPRs associated with the bony simulation. Overall, despite the differences in artifact magnitude, the results from both simulations confirm that the black hole pattern is largely caused by scatter originating from the patient tabletop.

III.B. Phantom studies

Figure 12 shows pelvis phantom scatter profiles for a lateral projection taken at a source angle of 8.1°. SKS, fASKS, and hybrid kernel estimates are compared against one another and against the measured scatter profile obtained using the slit-scan technique. For this highly impacted projection, the SKS kernels overestimate scatter by up to 25%, while the fASKS kernels underestimate scatter by up to 33%. The hybrid kernel algorithm produces a substantially more accurate estimate with a maximum error of less than 10%.

Figure 13 shows primary error sinograms that indicate the effectiveness of the different corrections over the entire angular range of the scan. The intensities of each sinogram row represent the primary error, averaged across the central 46 rows of the detector, for one projection angle. In the uncorrected sinogram [Fig. 13(a)], the average primary error magnitude is 71%, while the maximum error of over 250% occurs at the lateral projection angles for which the tabletop is in the FOV. The symmetric SKS algorithm markedly reduces this error; however, due to scatter overcorrection, the primary signal is now underestimated by up to 57% in the worst case [Fig. 13(b)]. The adaptive fASKS correction [Fig. 13(c)] shows further improvements over the bulk of the scan range. However, for the lateral views ranging from −30° to 30°, the primary signal is overestimated by 67% due to tabletop scatter. Use of the hybrid kernel method reduces the primary error to less than 10% across the entire scan range [Fig. 13(d)]. Figure 13(e) plots the average primary errors as a function of projection angle for each the corrections. For the lateral views, the hybrid kernel algorithm reduces the maximum magnitude of the primary error by ten times compared to fASKS and by four times compared to SKS.

FIG. 10. Soft tissue elliptical phantom simulation results. Due to the display level/window of 0/400 HU, the tabletop is not visible. (a) Primary-only reconstruction shows good HU uniformity and accuracy. (b) Primary plus object scatter reconstruction demonstrates more typical scatter-induced cupping artifacts with nonuniformities that are largely circularly symmetric. (c) Primary plus tabletop scatter reconstruction shows a severe bright-to-dark shading artifact in the vertical direction through the isocenter. (d) Average vertical profile through the ROI in (c) demonstrates that the magnitude of the black hole artifact exceeds 700 HU.

The reconstructed pelvis phantom images are displayed in Fig. 14. The uncorrected image [Fig. 14(a)] contains severe cupping artifacts and CT number inaccuracies that measure over 250 HU. The dark annulus located in the outer portion of the FOV is caused by the thick part of the bowtie filter which reduces the primary signal relative to the (more constant) scatter signal, thus increasing the local SPR. The slit-scan corrected image [Fig. 14(b)] has good uniformity except for the 30 HU (3%) band located toward the periphery. Also known as the “radar” artifact, this annular band is mainly caused by charge trapping and lag in the detector.^{31,32} The SKS-corrected image [Fig. 13(c)] is domed due to scatter overcorrection. When subtracted from the reference image [Fig. 13(d)], a 220 HU nonuniformity is measured through a vertical (anterior/posterior) profile [Fig. 13(e)]. Similarly, the fASKS correction results in a 200 HU nonuniformity caused by the black hole artifact [Fig. 13(h)]. After application of the hybrid kernel correction, the black hole artifact is largely eliminated [Fig. 13(i)] and the image uniformity error is reduced to approximately 50 HU excluding noise [Fig. 13(k)].

III.C. *In vivo* studies

Figure 15 shows averaged scatter profiles for a lateral projection (angle = -6.6°) from an Acuity pelvis scan of a large patient. As shown in Fig. 15(b), the maximum measured SPR is above 7 for this projection. Despite these chal-

lenging circumstances, the hybrid kernel algorithm estimates the scatter profile to within 10% of its measured value, thus providing significant improvements over the fASKS algorithm which underestimates scatter by more than 50% and over the SKS algorithm which overestimates scatter by 35%, so that it exceeds the measured signal.

Reconstructed axial images from the pelvis scan are shown in Fig. 16. Similar to the phantom results, the uncorrected image [Fig. 16(a)] has a dark outer circular band, as well as a black hole pattern. The SKS-corrected image [Fig. 16(b)] is domed, yet still with the black hole artifact. The fASKS-corrected image [Fig. 16(c)] is of relatively high quality except for the black hole. After application of the hybrid kernel correction, the black hole artifact is reduced by 60% from 245 to 98 HU and the resulting image [Fig. 16(d)] is improved to the point where it is comparable or superior to the slit-corrected reference image [Fig. 16(e)]. Visibility within the prostate region is substantially increased and overall uniformity is enhanced. Residual artifacts that remain may not all be necessarily due to scatter and could be attributed to beam-hardening effects, detector nonlinearities, charge trapping, or patient motion that occurs during the course of the 1 min acquisition.

Figure 17 shows reconstructions of the *in vivo* pelvis scan taken with a Varian OBI system before [Fig. 17(a)] and after [Fig. 17(b)] scatter correction using the hybrid kernel algo-

FIG. 11. Simulation results from the soft tissue ellipse with cortical bones. Due to the display level/window of 0/400 HU, the tabletop is not visible. (a) Primary-only reconstruction shows no artifacts. (b) Primary plus object scatter reconstruction shows typical scatter-induced cupping artifacts. (c) Primary plus tabletop scatter reconstruction shows a severe vertical bright-to-dark shading artifact. (d) Average vertical profile through the ROI in (c) shows that the magnitude of the black hole artifact exceeds 1000 HU (i.e., +400 to -600 HU).

rithm. Similar to the Acuity images shown in Fig. 16, the axial image reconstructed without any scatter correction has a dark outer rim as well as the black hole artifact. After application of the hybrid SKS algorithm, image quality

FIG. 12. Measured and estimated scatter profiles for the pelvis phantom (projection angle=8.1°). The hybrid kernel method produces scatter estimates that are within 10% of the measured scatter profile, while the symmetric (SKS) and adaptive (fASKS) methods either overestimate or underestimate scatter substantially. Each profile is the average of the individual profiles from the 46 central rows of the detector.

across the FOV is greatly improved and the black hole artifact, as measured by the difference in two circular ROIs shown, is reduced from 147 to 40 HU, representing an improvement of 73%. For comparison, diagnostic quality helical CT images from the same patient are shown in Fig. 17(c). While overall CBCT image quality does not match that of the helical CT image, it is seen that comparable CT numbers and image uniformities are attained after hybrid kernel correction. The line profiles taken through the sagittal slices demonstrate a 150 HU reduction in the vertical bright-to-dark nonuniformity across the isocenter with the resulting profile having a similar shape to that of the diagnostic CT profile. Residual artifacts that remain, as with the Acuity images shown in Fig. 16, may be due in part to nonscatter-related causes. Of note is that the treatment planning CT images and CBCT images were taken on different days and it is possible to see that the amount of air in the rectum has changed, which may be clinically significant.

IV. DISCUSSION

Patient tables used in radiation therapy must be mechanically stable and conducive for patient setup with and without immobilization devices. Consequently, the tabletops tend to be flatter, wider, and have higher attenuations than those used in diagnostic CT. We have shown that these properties can negatively impact onboard CBCT imaging performance. Specifically, tabletop scatter leads to a cupping artifact, re-

FIG. 13. Primary error sinograms (%) for the pelvis phantom with (a) no correction, (b) symmetric kernel correction (SKS), (c) adaptive kernel correction (fASKS), and (d) hybrid kernel correction. (e) Average primary errors plotted as a function of projection angle. The hybrid kernel correction produces superior primary estimates over the entire scan range. (All images share the same color bar.)

ferred to as the black hole, situated below the scan isocenter when data are acquired in offset-detector mode. Our results showed that the tabletop increased the total SPR by at least a factor of 2 in the most affected lateral views and that a maximum SPR of 7 was measured in an *in vivo* pelvis scan. Although some primary attenuation properties of radiation therapy tables have recently been investigated,³³ we believe this to be the first work to recognize the importance of scattered radiation from the patient table and its impact on image quality.

To address both tabletop scatter and object scatter, we developed a hybrid SKS correction algorithm that incorpo-

rates slab-based symmetric kernels and object-dependent adaptive kernels. Recently, we have shown that the adaptive kernels, generated by perturbing the shapes of the symmetric kernels according to local thickness variations across the projection, produce improved scatter estimates for elliptical and bodylike objects when compared to the estimates produced by the symmetric kernels. However, because the slablike tabletop extends beyond the patient in the lateral direction, the resulting scatter profile is better modeled by the symmetric kernels. By combining the adaptive and symmetric kernels into one algorithm, we were able to model both object and tabletop scatter. *In vivo* and phantom results show that

FIG. 14. Pelvis phantom CBCT reconstructions. All images except (b) were taken with the tabletop. (a) No scatter correction. (b) Reference image obtained using slit-scan scatter correction. (c) SKS correction. (d) The difference between (c) and (b). (e) Averaged vertical line profile through the ROI in (d). (f) fASKS correction. (g) The difference between (f) and (b). (h) Vertical line profile of (g). (i) Hybrid kernel correction. (j) The difference between (i) and (b). (k) Vertical line profile of (j). The profile measurements show that the SKS and fASKS corrections lead to 220 and 200 HU vertical nonuniformities, respectively. This error is reduced to 50 HU, excluding noise, after application of the hybrid kernel correction. Display window=450 HU (250 HU for the difference images).

the approach greatly reduces the black hole artifact and improves overall CBCT image quality even in very high SPR environments. The *in vivo* OBI results suggest that CBCT image quality may eventually be able to approach that of diagnostic CT if residual artifacts, stemming in part from detector charge trapping, patient motion, detector nonlineari-

ties, and remaining scatter estimation errors can be addressed.

To address the residual scatter estimation errors, one area for investigation concerns the weighting function w which, while tabletop-specific, was assumed to be independent of object size and shape. Moreover, a single weight was used

FIG. 15. (a) Estimated and measured scatter profiles for a lateral projection from the Acuity pelvis scan of a large patient. The displayed profiles are the average of the individual profiles from the 46 central rows of the detector for the projection angle = -6.6° . Scatter is greatly overestimated using the (symmetric) SKS algorithm and underestimated using fASKS, while the hybrid kernel model scatter estimate matches the measured scatter profile to within 10%. (b) The measured SPR across the projection is shown to exceed 7.

for the entire portion of the projection shadowed by the tabletop. A better estimate may be obtained by using a more complex weighting function that is both spatially and object-size-dependent. Another approach for improving the accuracy of the tabletop scatter correction algorithm is to use a data consistency criterion which states that two rays 180° apart from each other that pass through the isocenter must have the same primary signal. For opposing offset-detector lateral views, one of these readings will be affected by tabletop scatter and the other will not. We speculate that information gained from these two readings can be used to help augment the hybrid algorithm. Another possibility for improvement is to combine kernels derived from different materials such as water and polycarbonate to better model scatter from the object and the tabletop.

The influence of bony structures on measured scatter profiles was investigated via simulation. Because primary radiation is more highly attenuated by bone, the maximum object-induced and tabletop-induced SPRs increased, which also increased the intensity of the black hole artifact. No comparative evaluations of the effectiveness of the hybrid algorithm as a function of bony content were performed. However, since the estimated scatter profiles from a wide variety of phantom and *in vivo* data had good accuracy, the algorithm is implied to be fairly immune to, among other things, material heterogeneities. Thus, the detected scatter profiles, which are inherently of low spatial frequency, may only be moderately influenced by bones which are inherently of higher spatial frequency content. Nevertheless, since both the adaptive and conventional kernels are originally derived as-

FIG. 16. *In vivo* pelvis CBCT reconstructions from the Acuity scan. (a) No scatter correction. (b) SKS correction. (c) fASKS correction. (d) Hybrid kernel correction. (e) Reference correction using the slit-scan method. The CT number differences between ROIs 1 and 2 for these five images are 143, 146, 245, 98, and 160 HU, respectively. The hybrid kernel correction (d) yields the most uniform image and offers the highest visibility of the prostate and rectum anatomy. All images were beam-hardening corrected. Window = 450 HU.

FIG. 17. Comparison of CBCT and helical CT pelvis images. (a) CBCT reconstruction from the OBI data set with no scatter correction applied ("No SC"). (b) CBCT reconstruction after application of the hybrid kernel correction. (c) Diagnostic quality helical CT images from the same patient. Image nonuniformities, indicated by the differences between the two ROIs in each slice, are 147, 40, and 4 HU, respectively. The bottom row contains sagittal cuts taken through the plane marked by the arrow in the axial slices. (d) Vertical profiles taken through the marked rectangular region in each sagittal image show that the hybrid kernel correction yields an image with a similar HU profile to that from the helical (diagnostic) CT scan. Note that due to inexact registration, the marked region in the helical CT image does not exactly match that of the CBCT images. Image display window=450 HU.

suming complete soft tissue uniformity, there is a likely benefit to researching the effects of anatomic heterogeneities on detected scatter. Such efforts were beyond the scope of this work.

Patient tables used in linacs have various designs and configurations. We showed results from three tabletops of similar design using a different set of parameters for each model. The parameters were determined using a calibration process that is efficient and is required once per tabletop. Of note is that the parameters for these three models did not differ significantly. Thus, it may be possible to formulate an algorithm for determining weighting parameter values based on knowledge of tabletop shape and composition without necessarily undertaking a calibration procedure. It is also important to note that, having recognized patient table scatter as an important influence in CBCT image quality, it may be worthwhile to consider the design of new tabletops that produce less detected scatter.

The radiation treatment system reports the 3-D position of the tabletop with high accuracy and our algorithm combines use of this positional information with *a priori* knowledge of tabletop shape to generate a binary table mask indicating the location of the tabletop shadow in each projection. However, one assumption in our model that may still need to be investigated concerns how positional changes of the tabletop can affect scatter kernel properties. For example, for our experiments, the tabletop was centered in the left-right direction relative to the scan isocenter. However, if the tabletop is shifted to the left or right, it may be necessary to update the scatter kernel amplitudes and shapes to reflect that the tabletop position relative to the detector (i.e., the air gap) has changed for certain crucial lateral views. This aspect of kernel design was not explored.

Finally, it has been observed that the patient table-related scatter artifacts are largely manifested in the offset-detector

rather than the centered-detector mode. We attribute this partly to the higher SPRs associated with offset-detector scans because larger-sized objects are imaged. In addition, all lateral projections in centered-detector scans have the tabletop in the field-of-view. Thus, the resulting cupping artifact is more symmetric and potentially easier to address using conventional correction techniques. Nevertheless, the effects of tabletop scatter in centered-detector reconstructions may eventually prove to be significant, especially as overall CBCT image quality continues to improve and wider flat panels are developed allowing for larger transaxial fields-of-view to be achieved.

V. CONCLUSIONS

A distinct CBCT image artifact caused by radiotherapy patient tables is first identified. The artifact is traced to the scatter produced by the patient tabletop, and an efficient hybrid SKS algorithm is proposed to address the problem. Scatter from both the object-of-interest and the patient table is accurately estimated and image uniformities and HU accuracies in the resulting reconstructions are significantly improved. Future studies will be conducted to determine how much better suited the resulting images are to image-guided radiotherapy applications.

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