The influence of antiscatter grids on soft-tissue detectability in cone-beam computed tomography with flat-panel detectors

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The influence of antiscatter x-ray grids on image quality in cone-beam computed tomography (CT) is evaluated through broad experimental investigation for various anatomical sites (head and body), scatter conditions (scatter-to-primary ratio (SPR) ranging from ~10% to 150%), patient dose, and spatial resolution in three-dimensional reconstructions. Studies involved linear grids in combination with a flat-panel imager on a system for kilovoltage cone-beam CT imaging and guidance of radiation therapy. Grids were found to be effective in reducing x-ray scatter “cupping” artifacts, with heavier grids providing increased image uniformity. The system was highly robust against ring artifacts that might arise in CT reconstructions as a result of gridline shadows in the projection data. The influence of grids on soft-tissue detectability was evaluated quantitatively in terms of absolute contrast, voxel noise, and contrast-to-noise ratio (CNR) in cone-beam CT reconstructions of 16 cm “head” and 32 cm “body” cylindrical phantoms. Imaging performance was investigated qualitatively in observer preference tests based on patient images (pelvis, abdomen, and head-and-neck sites) acquired with and without antiscatter grids. The results suggest that although grids reduce scatter artifacts and improve subject contrast, there is little strong motivation for the use of grids in cone-beam CT in terms of CNR and overall image quality under most circumstances. The results highlight the tradeoffs in contrast and noise imparted by grids, showing improved image quality with grids only under specific conditions of high x-ray scatter (SPR > 100%), high imaging dose (D_{center} > 2 \text{cGy}), and low spatial resolution (voxel size \geq 1 \text{mm}). © 2004 American Association of Physicists in Medicine. [DOI: 10.1118/1.1819789]

Key words: flat-panel imager, computed tomography, cone-beam CT, imaging performance, x-ray scatter, artifacts, antiscatter grid, scatter correction, image-guided procedures

I. INTRODUCTION

Among the promising advanced applications of active matrix flat-panel imagers (FPIs) is cone-beam computed tomography (CT), providing the potential for combined soft-tissue visualization and submillimeter three-dimensional (3D) spatial resolution.1-3 Among the physical factors that challenge imaging performance in cone-beam CT are detector performance,4 dynamic range,5 and x-ray scatter—particularly the last, since the intensity of scattered x rays impinging on the detector can far exceed that of primary x rays.6 X-ray scatter simultaneously reduces the contrast of soft-tissue structures, increases image noise (i.e., reduces detectable quantum efficiency),7 and introduces streak and non-uniformity artifacts in 3D reconstructions.8 The development and implementation of strategies for management of x-ray scatter in cone-beam CT poses an area of research crucial to the deployment of this technology in diagnostic and image-guided procedures.

Numerous strategies for scatter management have been reported or are under investigation, combining scatter-reduction techniques from conventional transaxial CT and projection radiography, including: (1) knowledgeable selection of imaging geometry (air gap),7,9-12 (2) limitation of field size to the region of interest; (3) scatter-correction algorithms;13-15 and (4) use of an antiscatter x-ray grid on the surface of the detector.16-18 The last is a well-known, effective means of scatter rejection in projection radiography. Grids may be used in combination with an air gap, and, depending on scatter magnitude and detector performance,

may necessitate increased imaging dose [described by the Bucky factor (BF)] due to absorption of primary x rays in the vanes of the grid. Heavier grids [i.e., higher grid ratio (GR)] traditionally involve a higher Bucky factor (e.g., a quick rule of thumb: \( BF = 1 + GR/3 \) suggested by some reported clinical techniques). The need to increase dose by a Bucky factor with digital detectors is unclear, particularly for configurations featuring high detector quantum efficiency and input-quantum-limited performance. Absorption of primary x rays in grid vanes imparts a loss in noise-equivalent quanta (NEQ), but the conditions for which such degradation in NEQ significantly degrades detectability is unknown. Less clear still is whether image quality in cone-beam CT reconstructions is significantly degraded due to ring artifacts arising from subtle gridline shadows in the projection data which may persist even after dark-flood processing. The desire for low-dose imaging and the apprehension with regard to gridline artifacts have somewhat reduced the enthusiasm for incorporation of grids in early cone-beam CT systems. Furthermore, there is somewhat of a disparity among preliminary findings and recommendations (largely in private communication and at scientific meetings) as to the possible benefit or detriment associated with grids in cone-beam CT.

This work reports on a fairly comprehensive experimental investigation of the influence of antiscatter x-ray grids on image quality in cone-beam CT. Measurements cover a broad range of grid ratios, object size (head and body sites), scatter magnitude, imaging dose, and spatial resolution in cone-beam CT reconstructions. The results include the influence of grids on scatter artifacts (cupping and streaks), the susceptibility (or lack thereof) to gridline artifacts, the soft-tissue contrast, noise, and contrast-to-noise ratio (CNR), and the overall effect on image quality in patient images acquired under preclinical protocols. While the results are specific to a single cone-beam CT system (with a fixed detector type, pixel size, imaging geometry, etc.), the work provides insight into the factors governing imaging performance in cone-beam CT with grids and provides some indication regarding the circumstances and imaging tasks for which incorporation of a grid is of benefit to 3D image quality.

II. METHODS AND MATERIALS

A. Experimental conditions

1. Cone-beam CT on a medical linear accelerator

The imaging platform for all studies was a linear accelerator (Synergy RP, Elekta Oncology Systems) that incorporates an x-ray tube and flat-panel imager on the gantry as shown in Fig. 1(a). The system was designed for online radiographic, fluoroscopic, and cone-beam CT guidance of radiation therapy. The gantry rotates under computer control (up to 360° in 60 s), with the mechanical flex in source and detector components calibrated using a simple BB flex map and found to be geometrically reproducible to within subpixel precision. The x-ray tube (Dunlee PX 1402; 300 kHU, 1.0 mm focal spot, 15° anode angle) was operated at 120 kVp (2 mm Al+0.1 mm Cu added filtration) in pulsed fluoroscopic mode (1 pulse/frame; nominally 100 mA, 10 ms exposures). The longitudinal field of view (FOVz) at isocenter was varied from 2 to 25 cm through manual adjustment of the z-collimators. Axial field of view (FOVxy) was fixed at 26 cm for all cases, providing \( 26 \times 26 \text{ cm}^2 \) or \( \sim 40 \times 40 \text{ cm}^2 \) axial reconstructions in “centered” and “offset” detector geometries, respectively, below.

The FPI (PerkinElmer RID-1640A) is mounted on a ro-
botic arm opposite the x-ray tube and retracts flat against the gantry when not in use. The detector is a 1024 × 1024 (~41 × 41 cm$^2$) active matrix array of α-Si:H photodiodes and thin-film transistors at 400 μm pixel pitch, with 80% fill factor and a 133 mg/cm$^2$ Gd$_2$O$_2$S:Tb x-ray converter. Maximum frame rate is 3.5 fps, but for all measurements reported below, the device was triggered externally at 2.7 fps, providing ~320 projections across 360° in 120 s.

Cone-beam CT image acquisition, processing, and reconstruction were performed using a system developed in our laboratory. The acquisition computer (Dual Xeon Intel CPU, 2.80 GHz, 2 GB RAM) synchronizes x-ray tube exposure and FPI readout during gantry rotation. Gantry angle is monitored to within ±0.1° at each projection view by means of analog input to the linac control system. Projections were corrected for stationary variations in offset and gain by the mean of 50 “dark” and “flood” fields acquired in the absence and presence, respectively, of flat-field x-ray exposure. A simple geometric calibration was performed as described previously and similar to that of Fahrig and Holdsworth in which the position of the “piercing point” (i.e., the intersection of the ray containing the source and isocenter with the detector plane) is measured as a function of gantry angle from projections of a steel BB placed at isocenter. To avoid hysteresis effects, calibration and acquisition were always performed under clockwise rotation. Deviations from an ideal circular orbit are corrected by means of small in-plane shifts of the projection matrix according to the calibration map. While this simple geometric correction does not account for a variety of geometric nonidealities (e.g., detector tilt), it has shown to provide correction of the most significant effects; a more sophisticated technique that measures and corrects for all geometric nonideality is under development.

The system has a source-to-isocenter distance of ~100 cm and a source-to-detector distance of ~153 cm. With the FPI centered with respect to the central axis of the x-ray beam, this gives a ~26 × 26 × 26 cm$^3$ reconstruction FOV at isocenter, just large enough to encompass a human head without serious projection truncation. Alternatively, the detector is offset laterally by ~10 cm (or up to ~20 cm) to provide ~40 × 40 × 26 cm$^3$ volumetric FOV from projections acquired over 360°, sufficient to cover fairly large anatomical sites with minimal projection truncation.

At full resolution (i.e., 400 μm pixels at the detector plane; magnification ~1.5) the voxel size, $a_{\text{vox}}$, in cone-beam CT reconstructions is 0.25 mm in $(x,y,z)$. Due to blur in the x-ray conversion phosphor and application of a smooth (Hanning) filter during reconstruction, the system spatial resolution is ~0.6–0.7 mm (determined from the full width at half maximum in images of thin wire). Therefore, projection data are typically processed to “half-resolution” or “quarter-resolution,” giving $a_{\text{vox}}=0.5$ or 1.0 mm, respectively.

Volume image reconstruction was performed on either the acquisition PC or a remote “reconstruction server” (Quad Intel Xeon CPU, 2.0 GHz, 4 GB RAM) using a modified Feldkamp algorithm for cone-beam reconstruction developed in research collaboration (see Acknowledgments). Reconstruction of a quarter-resolution 256$^3$ volume (or a half-resolution 512$^3$ volume) required 1 min, 33 s (or 12 min, 7 s) on the acquisition PC.

Imaging dose was determined using a calibrated ion chamber and 16 and 32 cm cylindrical dosimetry phantoms. Dose was varied through adjustment of tube current (50–100 mA) and exposure time (10–40 ms). As listed in Table I, measurements were performed at one of four dose levels, denoted: D1 (50 mA; 10 ms; giving 0.5 mAs/projection and ~160 total mAs/scan); D2 (100 mA; 10 ms; giving 1 mAs/projection and ~320 total mAs/scan); D3 (100 mA; 20 ms; giving 2 mAs/projection and ~640 total mAs/scan); and D4 (100 mA; 40 ms; giving

### Table I. Summary of experimental parameters examined in investigating the influence of antiscatter grids on soft-tissue imaging performance.

<table>
<thead>
<tr>
<th>Experimental parameter</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Imaging phantom</td>
<td>16 cm diameter Head cylinder</td>
</tr>
<tr>
<td></td>
<td>32 cm diameter Body cylinder</td>
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<tr>
<td>Tissue-equivalent inserts</td>
<td>Solid water background (0 HU)</td>
</tr>
<tr>
<td></td>
<td>Adipose AP6 (~100 HU)</td>
</tr>
<tr>
<td></td>
<td>Breast BR12 (~50 HU)</td>
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<td></td>
<td>Brain SR2 (8 HU)</td>
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<tr>
<td></td>
<td>Liver L1 (85 HU)</td>
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<tr>
<td></td>
<td>Teflon (~1000 HU)</td>
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<tr>
<td>Grids</td>
<td>No grid</td>
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<tr>
<td></td>
<td>GR=6:1 (85 lpi)</td>
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<tr>
<td></td>
<td>GR=6:1 (103 lpi)</td>
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<tr>
<td></td>
<td>GR=8:1 (103 lpi)</td>
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<td></td>
<td>GR=10:1 (103 lpi)</td>
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<tr>
<td></td>
<td>GR=12:1 (103 lpi)</td>
</tr>
<tr>
<td>Grid orientation</td>
<td>Along-ψ: grid lines perpendicular to axis of rotation</td>
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<tr>
<td></td>
<td>Along-ψ: grid lines parallel to axis of rotation</td>
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<tr>
<td></td>
<td>45°: grid placed at 45° to detector matrix</td>
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<tr>
<td></td>
<td>Small angle: grid placed at ~5° relative to Along-ψ</td>
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<tr>
<td>Detector geometry</td>
<td>Centered (detector centered on central ray; FOV$_{xy}$ ~26 cm)</td>
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<tr>
<td></td>
<td>Offset (detector translated ~10 cm in ψ; FOV$_{xy}$ ~40 cm)</td>
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<tr>
<td>Longitudinal FOV$_z$</td>
<td>FOV$<em>z$ = 2 cm ($\phi</em>{\text{cone}}=1.1\degree$)</td>
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<tr>
<td></td>
<td>FOV$<em>z$ = 10 cm ($\phi</em>{\text{cone}}=5.7\degree$)</td>
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<tr>
<td></td>
<td>FOV$<em>z$ = 20 cm ($\phi</em>{\text{cone}}=14.8\degree$)</td>
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<tr>
<td>Imaging dose</td>
<td>D1(0.5 mAs/proj):</td>
</tr>
<tr>
<td></td>
<td>$D^\text{proj}_{\text{center}}=0.8 \text{ cGy}$</td>
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<tr>
<td></td>
<td>D2(1.0 mAs/proj):</td>
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<tr>
<td></td>
<td>$D^\text{proj}_{\text{center}}=1.6 \text{ cGy}$</td>
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<td></td>
<td>D3(2.0 mAs/proj):</td>
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<tr>
<td></td>
<td>$D^\text{proj}_{\text{center}}=3.2 \text{ cGy}$</td>
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<tr>
<td></td>
<td>D4(4.0 mAs/proj):</td>
</tr>
<tr>
<td></td>
<td>$D^\text{proj}_{\text{center}}=6.4 \text{ cGy}$</td>
</tr>
<tr>
<td>Voxel size</td>
<td>Full-resolution: $a_{\text{vox}}=0.25 \text{ mm}$</td>
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<td></td>
<td>Half-resolution: $a_{\text{vox}}=0.5 \text{ mm}$</td>
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<tr>
<td></td>
<td>Quarter-resolution: $a_{\text{vox}}=1.0 \text{ mm}$</td>
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4 mAs/projection and ~1280 total mAs/scan). From the ion chamber measurements, the relation to total scan dose at the center of a 32 cm Body phantom is 0.0034 cGy/total mAs, giving: D1 (0.6 cGy); D2 (1.1 cGy); D3 (2.2 cGy); and D4 (4.4 cGy). Similarly, the total scan dose to water at the center of a 16 cm Head phantom is 0.005 cGy/total mAs, giving: D1 (0.8 cGy); D2 (1.6 cGy); D3 (3.2 cGy); and D4 (6.4 cGy). These values represent dose to water for “centered” detector geometry (the nominal configuration below) and largest FOVz, (25 cm; worst case). Results below are reported in terms of mAs/projection, which is a convenient metric from the image acquisition standpoint. The values above and in Table I relate mAs/projection to total imaging dose.

2. Antiscatter grids

A variety of antiscatter grids were employed to investigate the effect of GR and grid lines-per-inch (lpi) on soft-tissue imaging performance. Grid ratios of 6:1, 8:1, 10:1, and 12:1 (each at 103 lpi) were investigated. In addition, grids at 85 and 103 lpi were examined (GR=6:1). All grids (Soxey Products linear grids, Korea) were new and in excellent condition, verified by exposure to direct-conversion film by the manufacturer. The grids comprised linear Pb vanes with Al interspacers, with focal length 102–183 cm (consistent with the 155 cm source-to-detector distance). The transmission factor for each grid was measured using a silicon diode (R100 detector with Barracuda exposure meter; RTI Electronics, Molndal Sweden) stepped through 40 positions behind the grid (0.025 mm increments, ten measurements at each position) on a motorized translation stage. Mean transmission factors were comparable in each case: (0.70, 0.69, 0.66, 0.65, and 0.62) ±0.01 for the 6:1 coarse, 6:1 fine, 8:1, 10:1, and 12:1 grids, respectively. Each grid was 15 ×18 in. (38.1 ×45.7 cm²), which was smaller in one dimension than the FPI, but as shown in Fig. 1(b), was sufficient to cover the lateral extent of the detector, which is the pertinent dimension for most of the studies reported below. In all cases, the grid was placed in direct contact with the cover plate of the FPI, which is ~9 mm from the surface of the x-ray converter. A “bucky” system (which might reduce the visibility of gridlines in projections by motion blur during exposures) was not used. Unless otherwise stated, the grid was in exactly the same position for flood-field corrections as for projection images.

3. Imaging phantoms

Two imaging phantoms formed the basis for the quantitative investigation of image artifacts and soft-tissue CNR reported below: a 16 cm diameter cylindrical “Head” phantom, and a 32 cm diameter cylindrical “Body” phantom. In each case, a 16 cm diameter solid water cylinder machined to accept a variety of tissue-equivalent inserts was used. The Head phantom is simply the solid water cylinder with tissue-equivalent inserts, while the Body phantom consisted of the solid water cylinder with inserts placed within a 32 cm acrylic annulus, as illustrated in Fig. 1(c).

Tissue-equivalent inserts (Gammex RMI, Madison WI) included “Adipose” (~100 HU), “Breast” (~50 HU), “Brain” (8 HU), “Liver” (85 HU), and two Teflon rods (~1000 HU) intended to simulate cortical bone, each set within the 16 cm solid water (0 HU) cylinder. A 1 cm diameter hole in the center of the cylinder contained a 0.127 mm steel wire for examination of the axial point-spread function in cone-beam CT reconstructions.

4. Imaging conditions: Scatter, dose, and spatial resolution

As summarized in Table I, experimental conditions were varied over a broad range of x-ray scatter magnitude, imaging dose, and spatial resolution to form a fairly comprehensive examination of the influence of grids on soft-tissue imaging performance. The magnitude of x-ray scatter was affected by selection of anatomical site (Head and Body phantoms) and longitudinal FOVz, (varied from 2 to 20 cm). Imaging dose was varied by adjustment of mAs per projection from 0.5 to 4 mAs per projection, as detailed above. Spatial resolution in cone-beam CT reconstructions was adjusted by increasing the voxel size from “full-resolution” (a_vox=0.25 mm) to half-resolution (a_vox=0.5 mm) and quarter-resolution (a_vox=1 mm) by application of an appropriate (2 ×2 or 4 ×4) pixel-averaging aperture. A Hanning reconstruction filter was used in all cases, believed to be a reasonable choice for the task of soft-tissue imaging.

B. Analysis of imaging performance

1. X-ray scatter artifacts: Cupping and streaks

As reported previously, the magnitude of x-ray scatter at the detector in cone-beam CT can be very high [e.g., scatter-to-primary ratio (SPR) exceeding 100%] and is determined primarily by imaging geometry (air gap), object size, and FOVz. The geometry of the system in Fig. 1(a) provides an air gap that is optimal with regard to scatter rejection, spatial resolution (geometric sharpness and focal spot blur), detector noise, etc. Still, SPR at the detector is high (e.g., ~40% for the Head and ~140% for the Body for large FOVz) and is known to cause significant image nonuniformity artifacts (cupping and streaks) in 3D reconstructions.

The magnitude of such artifacts was measured for the Head and Body phantoms as a function of FOVz. Signal profiles were extracted from axial reconstructions (11 rows averaged) through regions not containing tissue-equivalent inserts (solid water only). For the x-ray spectrum used, the attenuation coefficient of water is ~0.19 cm⁻¹, with x-ray scatter typically causing a reduction in voxel values. We quantify the CT number inaccuracy near the center of image reconstructions by a simple metric, \( t_{\text{cup}} \), defined in relation to the magnitude of the cupping artifact

\[
 t_{\text{cup}}(\%) = 100 \times \frac{\mu_{\text{edge}} - \mu_{\text{center}}}{\mu_{\text{edge}}},
\]

where the \( \mu \) refer to the average over ~1 cm² regions (21 ×21 voxels for half-resolution, 11 ×11 voxels for quarter-
resolution) near the edge or center of the phantom in an axial image. Scatter artifacts were evaluated quantitatively and qualitatively for each combination of grid and FOV.\(z\) (Table I).

2. Grid perturbation and orientation: Line artifacts and rings

Because even faint, stationary imperfections in projection data (e.g., defective lines or pixels) can cause significant artifacts in CT reconstructions (e.g., ring artifacts),\(^6\) it was essential to examine the influence of gridline shadow artifacts on reconstruction image quality. Concerns were threefold: (1) Imperfect flood-field correction, even for cases in which the grid is in place and undisturbed for both flood-field and projection data acquisition; (2) Perturbation (e.g., misalignment) of the grid between flood-field and projection data acquisition, causing gross errors in flood-field correction; (3) Grid orientation [viz., gridlines along detector rows \((u)\) or detector columns \((v)\); See Fig. 1(b)] for “centered” and “offset” detector geometries, where the last combination (“Along-\(v\)/offset”) is presumably a poor configuration, since the grid would not be focused on the x-ray source.

To examine the influence of gridline shadows on CT reconstructions, the Head and Body phantoms were imaged with the following five configurations of the 8:1 grid in both flood-field and projection data: No-grid; grid Along-\(u\), grid Along-\(v\); grid at 45° to detector matrix; and grid at a small angle (~5°) to the detector matrix. This gave 25 possible combinations of grid orientation between flood-field and projection data acquisition. The five combinations for which the orientation was identical between flood-fields and projection data (e.g., no-grid/no-grid; Along-\(u\)/Along-\(u\), etc.) provided examination of the first concern (imperfect flood-field correction, e.g., due to beam-hardening). The other 20 combinations (e.g., no-grid/Along-\(u\), etc.) represent fine and gross perturbations of the grid between flood-field and projection data acquisition, with significant gridline shadows expected in the processed projection data. Finally, images of the Body phantom were acquired in centered and offset detector geometries, with the 8:1 grid oriented Along-\(u\) and Along-\(v\), the four combinations providing quantitative and qualitative examination of image quality in each case.

3. Soft-tissue contrast-to-noise ratio

Contrast, noise, and CNR were evaluated in each tissue-equivalent insert in the Head and Body phantoms. In each case, a ~1 cm\(^2\) region of interest (ROI) was identified within the insert and immediately adjacent to the insert (at the same radius) in the solid water cylinder. The mean and standard deviation (\(\bar{\mu}\) and \(\sigma\), respectively) in voxel values in each ROI was computed, giving the absolute contrast (\((\bar{\mu}_{\text{tissue}} - \bar{\mu}_{\text{water}})\)), voxel noise \((\sigma_{\text{water}}\), in all cases \(\sim \sigma_{\text{tissue}}\)), and CNR:

\[
\text{CNR} = \frac{|\bar{\mu}_{\text{tissue}} - \bar{\mu}_{\text{water}}|}{\sigma_{\text{water}}}. \tag{2}
\]

CNR was evaluated in consecutive axial slices covering ~1 cm about the central plane (20 slices for half-resolution, 10 slices for quarter-resolution), with the standard deviation in CNR between slices providing the experimental error estimate (error bars). Measurements were performed for all iterations of the independent variables considered: grid ratio, FOV,\(z\), dose, and voxel size (Table I).

4. Patient imaging

Following analysis of image artifacts and soft-tissue CNR in the Body and Head phantoms, cone-beam CT images of patients undergoing radiation therapy were acquired under site-specific protocols for preclinical investigation. Patient images were acquired on the system shown in Fig. 1(a) using a 10:1 grid at various FOV.\(z\) (10 and 22–25 cm) and voxel size (\(a_{\text{vox}}=0.5\) and 1.0 mm). A total of 30 scans were performed across eight patients, with sites including pelvis, abdomen, and head-and-neck. All scans were performed using offset-detector geometry, with ~320 projections acquired at 120 kVp (2 mm Al+0.1 mm Cu added filtration) and 1.3 mAs/projection across 360°. All subjects were patients undergoing radiation therapy at our institution, and the imaging studies were performed under informed consent and with ethics approval. Simple preference tests (grid versus no-grid) were conducted using three radiation oncologists and three physicists as observers. Images were similarly window/leveled about each image histogram to allow fair comparison, and observers were shown axial and coronal views in side-by-side comparison between grid and no-grid cases. Observers were asked to rate which case allowed better soft-tissue visibility and to provide qualitative rationale for their preference in terms of image artifact, contrast, noise, and spatial resolution.

III. RESULTS

A. X-ray scatter artifacts: Cupping and streaks

The degree to which grids reduce cupping and streak artifacts associated with x-ray scatter is shown in Fig. 2. For small FOV.\(z\) (2 cm, corresponding to SPR ~10% for the Head phantom),\(^6\) cupping artifacts are almost negligible \((t_{\text{cup}} \sim 3.7\%)\), and as shown by the image profiles in Fig. 2(a), images with and without grids are equivalent. At higher FOV.\(z\) (20 cm, corresponding to SPR ~30%–40% for the Head phantom),\(^6\) cupping and streak artifacts are more significant. As shown in Fig. 2(b), the cupping artifact for the Head phantom is \(t_{\text{cup}} \sim 10\%\) without a grid and reduces in proportion to grid ratio. For example, addition of a heavy (12:1) grid reduced the artifact to \(t_{\text{cup}} \sim 4\%)\), which is about the same level of artifact as in the low-scatter (FOV.\(z\) = 2 cm) case. Therefore, grids appear effective in reducing cupping artifacts, which can be a confounding issue in cone-beam CT, particularly regarding CT number accuracy and display dynamic range.

The influence of grids on such artifacts under even higher-scatter conditions is shown in Fig. 3, in which axial images of the Body phantom are shown as a function of grid ratio and FOV.\(z\). Without a grid (first column), images are strongly degraded by scatter with increasing FOV.\(z\). Although there is
a truncation artifact (increased voxel values) evident at the periphery of each image due to the phantom size and imaging geometry, the increase in cupping as \( \text{FOV}_z \) is varied from 2 to 20 cm is attributable to x-ray scatter. Addition of a grid significantly reduces the cupping artifact. For small \( \text{FOV}_z \) (2 cm, corresponding to SPR \( \sim 30\% \) for the Body phantom), the slight cupping artifact is removed at the cost of increased image noise (investigated below in terms of image CNR). For large \( \text{FOV}_z \) (20 cm, corresponding to SPR \( \sim 140\% \) for the Body phantom), the strong cupping artifact is greatly reduced by even a light (6:1) grid, and improves monotonically with grid ratio.

**B. Grid perturbation and orientation: Line artifacts and rings**

The effect of gridline artifacts and perturbations to the grid between flood-field correction data and the projection image data is illustrated in Fig. 4. The top set of figures (a) shows magnified views (65 × 65 pixels) of dark-flood-corrected projection images, and the bottom set of figures (b) shows magnified views (169 × 169 voxels) of image reconstructions in the regions of the Breast insert in the 16 cm Head phantom. Each column of images shows cases in which the grid (GR=8:1) was oriented in the direction indicated for projection acquisition—e.g., gridlines along FPI rows \((u)\), along FPI columns \((v)\), at 45\( ^\circ \) to the detector matrix, etc. Each row of images shows cases in which the grid was oriented along the direction indicated in the preceding set of flood-field correction images. Therefore, the grid was identically oriented in flood-field and projection images along the diagonal, and off-diagonal cases represent mismatch or perturbation in grid orientation between flood-field and projection images.

Along the diagonal of Fig. 4(a), gridline “shadows” are removed fairly well from the projection image simply by dark-flood correction. Off-diagonal, however, where the grid is oriented differently between flood-field and projection images, gridline and shading artifacts are clearly evident in the processed projection data. The presence of such stationary artifacts in the processed projection data immediately raise concerns that cone-beam CT reconstructions would be seriously degraded by ring and shading artifacts. As seen in Fig. 4(b), however, the system is surprisingly robust against such artifacts in 3D reconstructions. Along the diagonal, reconstructions appear nearly equivalent (with slightly increased stochastic noise in all but the no-grid case, as discussed below in relation to CRN). Somewhat surprisingly, off-diagonal images exhibit little or no degradation. For example, consider the case in which the grid was oriented Along-\( u \) in projections and Along-\( v \) in flood-fields, or at 45\( ^\circ \) in projections and Along-\( u \) in flood-fields. In each of these cases, strong gridline and shading artifacts are evident in the processed projections; however, there is little or no degradation in the image quality of 3D reconstructions (verified quantitatively in terms of CRN). The worst case appears when the grid is perturbed by a “small angle” (\( \sim 5\% \)) between projection and flood-field sets, and deterministic dark ring artifacts appear in the reconstructed images at a level comparable to the quantum noise.

The robustness of 3D reconstructions against gridline artifacts was a somewhat surprising—and encouraging—result. On initiating the experiments, we had hypothesized a sensitivity to both fine and gross perturbations of the grid; however, the results largely disprove that hypothesis. Sensitivity to perturbations could pose a serious hurdle to practical implementation, requiring a greatly increased level of quality assurance to guarantee that flood-fields are up-to-date. Alternatively, one might implement a motion-bucky system to minimize gridlines in both projection and flood-field images. There is a distinct degradation for small angle perturbation of
the grid, so regular quality assurance is still likely to be important; however, overall the system was surprisingly robust against ring artifacts arising from gridline shadows. In light of the results, we hypothesize the following in regard to the lack of gridline ring artifacts in 3D reconstructions: The system shown in Fig. 1(a) employs a geometric calibration or “flex-map” to account for mechanical flex in the source and detector components through the 360° orbit. This technique has shown to provide a fairly reproducible calibration that is essential to achieving high image quality on a mechanically nonideal gantry, such as the linear accelerator. While the reproducibility of the calibration is high (better than 0.4 pixels, or ~0.1 mm), we hypothesize that subpixel imperfections in geometric calibration are sufficient to reduce the impact of gridline artifacts on 3D reconstruction. Effectively, imperfect correction of the two-dimensional (2D) coordinates in the detector plane with respect to the 3D coordinates of the image reconstruction constitute a “3D Bucky” that helps to “smear-out” the gridlines upon backprojection in a manner analogous to that by which a motion-bucky blurs gridlines during a single-shot exposure. Testing this hypothesis is the subject of future work, being carried out on a cone-beam CT imaging bench that offers near-ideal circular orbit and features source and detector translation stages that can be driven during acquisition to simulate the nonideal flex-map.

We also examined the effect of grid orientation in centered and offset source-detector geometries. Using the 32 cm Body phantom, the grid (GR=8:1) was aligned along \( \mu \) and along \( \nu \), as described above, with the detector in the centered and offset positions. The only case among these four combinations that conceptually poses a problem is the “Along-\( \nu \)/offset” combination, since the grid is not correctly focused on the x-ray source. (Rather, it is focused on a line in space offset laterally from the x-ray source.) Qualitative inspection of resulting Body phantom images (not shown for space considerations) and quantitative analysis of CNR indicated that there was little or no difference in imaging performance among the four cases (including, somewhat surprisingly, the worst case Along-\( \nu \)/offset) at any setting of FOV \( z \) and voxel size. Examination of the raw and processed projection images suggests that flood-field correction is effective in removing gridline shadows even in the worst case, although the worst case may breakdown (increased noise) at low dose. The main finding here is that the Along-\( \mu \) orientation (the default in all studies herein) was robust in both centered and offset geometries. That the Along-\( \nu \) orientation provided comparable results speaks to the robustness of the system against gridline artifacts and opens the possibility for 2D (rather than linear) grids.

C. Soft-tissue contrast-to-noise ratio

The CNR in various soft-tissue-equivalent structures (e.g., Breast, Adipose, Liver, etc.) was measured in 3D image reconstructions as a function of grid ratio (6 grids in Table I), body site (Head and Body phantoms), FOV \( z \) (2, 10, and 20 cm), dose (four settings in Table I), and voxel size \( (a_{vox} = 0.25, 0.5, \text{and } 1.0 \text{ mm}) \). The example results summarized in Figs. 5–8 illustrate quantitatively and qualitatively the conclusions with respect to grid selection for the task of soft-tissue discrimination.
Figure 5 plots soft-tissue CNR as a function of dose for the Adipose-equivalent insert in the 32 cm cylindrical Body phantom. The first row [Figs. 5(a) and 5(b)] corresponds to a voxel size of $a_{\text{vox}}=0.5$ mm (half-resolution), and the second row [Figs. 5(c) and 5(d)] to $a_{\text{vox}}=1.0$ mm (quarter-resolution). The first column corresponds to FOV$_z=2$ cm (low-scatter), and the second column to FOV$_z=20$ cm (high-scatter). For low-scatter conditions (FOV$_z=2$ cm), it is clear that the system without a grid provides superior CNR to any grid configuration at either spatial resolution. For high-scatter conditions (FOV$_z=20$ cm), grids begin to offer some advantage. While CNR is degraded with increasing FOV$_z$ in all cases, for the higher-resolution case ($a_{\text{vox}}=0.5$ mm), the no-grid system is still superior to all grid configurations. For the lower-resolution case ($a_{\text{vox}}=1.0$ mm), however, the degradation in CNR imparted by increased FOV$_z$ is more severe.
for the no-grid case than for the grids, and improved CNR is achieved with higher grid ratios, at least for doses above 2 mAs/projection. An understanding of the tradeoffs suggested here is gained by considering the factors that affect image contrast (GR and SPR) and noise (dose and spatial resolution) separately, as discussed below.

Figure 6 plots soft-tissue CNR as a function of imaging dose for the Breast-equivalent insert in the 16 cm cylindrical Head phantom. As in Fig. 5, results are shown for half-resolution and quarter-resolution and FOV = 20 cm, and similar trends are observed. For low-scatter, higher-resolution images, the no-grid system is superior to any grid configuration. The advantage is less clear, however, as FOV is increased (increased SPR) or spatial resolution is decreased (noise is reduced). For the high-scatter, higher-resolution case [Fig. 6(b)] and for the low-scatter, lower-resolution case [Fig. 6(c)] all systems perform nearly equivalently within experimental error. For the high-scatter, lower-resolution case [Fig. 6(d)], however, CNR improves with grid ratio. The influence of grids on contrast and noise separately is discussed below, showing that grids enjoy an advantage under conditions of high x-ray scatter, high imaging dose, and low spatial resolution.

Figure 7 shows the effect of grid selection on soft-tissue CNR for the highest-scatter case (FOV = 20 cm) in both Head and Body sites at half- and quarter-resolution. In each case, CNR is plotted as a function of grid ratio at three levels of imaging dose, with grid ratio 0 corresponding to no-grid. Four distinct possibilities are revealed: (1) For the Body imaged at half-resolution (vox = 0.5 mm) and any dose level considered, CNR monotonically reduces with GR; therefore, grids degrade CNR. (2) At quarter-resolution (vox = 1.0 mm), CNR is marginally improved for the high-dose case (4 mAs/projection); therefore, grids improve CNR given sufficient dose. (3) For the Head imaged at half-resolution (vox = 0.5 mm) and any dose level considered, CNR is the same for all GR; therefore, grids have no effect on CNR. (4) At quarter-resolution (vox = 1.0 mm), CNR improves monotonically with GR; therefore, grids improve CNR.

Taken together, the observations of Fig. 7 indicate that the influence of grids on imaging performance depends strongly on the imaging conditions (SPR and dose) and imaging task (visualization of soft-tissue structures at low resolution versus high-contrast structures at high resolution). Furthermore, these results encapsulate precisely the seeming disparity of (largely unpublished) findings and recommendations as to whether use of a grid is beneficial in cone-beam CT—some groups arguing against grids, some arguing that there is little or no effect, and some arguing in favor of grids. Given sufficient dose, in cases where spatial resolution is secondary to soft-tissue detectability (e.g., visualization of a subtle lesion, with vox = 1 mm), selection of a heavier grid will benefit performance of the imaging task. In cases where high spatial resolution is required (e.g., visualization of bony detail, with vox = 0.5 mm or finer), there is no advantage to using a grid. As discussed later, the influence of grids on image contrast and noise separately helps to shed light on the tradeoffs.
Also evident in Fig. 7 is a subtle difference in soft-tissue CNR for coarse and fine grids (the former plotted just below GR=6:1, and the latter just above). Close examination in each case at GR~6:1 reveals a slight, but experimentally reproducible, improvement for the fine grid (103 lpi) compared to the coarse grid (85 lpi). Qualitative examination of the fine- and coarse-grid images indicates that, similar to the differences among all grids as described below, the difference is not due to image artifact (e.g., rings), but to the contrast and stochastic noise in the image.

The dependence of soft-tissue detectability on grid selection, FOV, and spatial resolution, as quantified by the CNR in Figs. 5–7, is illustrated qualitatively in Fig. 8, showing that the relative improvement or degradation in image quality is fairly subtle. In each case, a magnified view in a ~3 × 3 cm² region about the Adipose-equivalent insert in the Body phantom is shown. Grayscale window is equivalent in each case. Each matrix shows the region of interest at varying FOV, and GR (0–12:1) and qualitatively demonstrates the tradeoff between improved contrast and increased noise imparted by antiscatter grids. In the higher-resolution case [a_\text{vox}=0.5 mm; Fig. 8(a)], the increase in contrast offered by higher GR is largely outweighed by increased noise in all cases. For low x-ray scatter (first row; FOVₓ=2 cm), grids are seen to degrade image quality. For high x-ray scatter (third row; FOVₓ=20 cm), the argument in favor of grids can begin to be appreciated: a clear increase in contrast is achieved through application of a grid; however, at fine voxel size, this benefit is largely outweighed by an increase in stochastic noise (the “dose penalty”), as quantified by the CNR in Fig. 7(a). In the lower-resolution case [a_\text{vox}=1 mm; Fig. 8(b)], the dose penalty incurred by heavier grids is overcome by reconstructing at a larger voxel size. For example, in the last row of Fig. 8(b) [FOVₓ=20 cm], the increased contrast imparted by grids is only marginally compromised by increased noise, resulting in a slight increase in image quality, evident in medium and high-dose cases of Figs. 5(d) and 7(b). Therefore, if the imaging task allows lower spatial resolution (a_\text{vox}=1 mm) and the clinical constraints tolerate increased dose (2 mAAs/projection or greater), then image CNR improves with application of a grid.

D. Patient images

Figures 9–12 show example patient images acquired with and without a 10:1 grid at various FOVₓ and spatial resolution (voxel size). As described above, these results were qualitatively examined by six observers in simple preference tests, rating preference of grid versus no-grid cases and noting the influence of contrast, noise, artifacts, and spatial resolution on their choice. A small subset of the total 60 image reconstructions are shown below, with qualitative observer preference results related to the quantitative results of Sec. III C.

Figures 9 and 10 illustrate axial and coronal images of the same patient in the region of the prostate, imaged with FOVₓ=22 cm (high-scatter conditions) and reconstructed at a_\text{vox}=0.5 mm (Fig. 9) and a_\text{vox}=1.0 mm (Fig. 10). In the former case (Fig. 9), observers unanimously preferred the no-grid image despite the presence of an obvious x-ray scatter shading artifact. Observers attributed preference to reduced noise and better definition of soft-tissue boundaries in the no-grid case. These results are generally consistent with the results of Figs. 5(b) and 7(a) that show reduction in CNR for higher spatial resolution reconstructions.
Figure 10 shows the same image data reconstructed at lower spatial resolution ($a_{\text{vox}}=1\,\text{mm}$), and observer preference was somewhat divided. Observers agreed that the axial image was superior with the grid, whereas the coronal image was superior for the no-grid case, presumably due to the influence of the shading artifact that is more pronounced in the axial view. Observers were divided evenly in terms of overall preference: those preferring the grid images noted less shading artifact and little influence of image noise on soft-tissue detectability; those preferring the no-grid images noted better delineation of the prostate, particularly in the coronal view. These results are somewhat consistent with the results of Figs. 5(d) and 7(b), where the influence of grids on CNR is weaker and more dependent on imaging dose. Of course, there is more to observer preference than just CNR, as noted in observers’ preference for axials versus coronals and tolerance of shading artifacts. Furthermore, observers may be able to “see through” a certain amount of correlated or uncorrelated image noise, in which case observers benefit more from improved contrast than they suffer from increased noise; therefore, CNR is a somewhat limited surrogate for human observer performance, though it is a useful touchstone for basic imaging performance.

Figure 11. Patient images acquired (a) without and (b) with a grid. FOV$^z=10\,\text{cm}$, and $a_{\text{vox}}=1.0\,\text{mm}$. For this lower-scatter case, observers preferred the no-grid images for delineation of the prostate and surrounding soft-tissue structures, noting reduced image noise, consistent with the results above that show improved CNR with grids only under high-scatter conditions.
Continuing with the lower-resolution case, Fig. 11 shows images of the same patient acquired at FOV$_z$=10 cm (moderate scatter conditions). Observers were unanimous in overall preference of the no-grid case, noting that shading artifacts were not a significant factor limiting soft-tissue detectability, and that the lower noise level in both axial and coronal views significantly improved soft-tissue delineation. This is generally consistent with the quantitative results above that suggest improvement in CNR with grids only under high-scatter conditions.

Finally, Fig. 12 shows images of a patient in the region of the head and neck acquired at FOV$_z$=25 cm and reconstructed at $a_{\text{vox}}=0.5$ mm (which is typical of our head and neck imaging protocol). Consistent with the results of Figs. 6(b) and 7(c), which show little or no effect of grids on CNR, observers were evenly divided between the grid and no-grid cases, noting little appreciable difference between the two in terms of soft-tissue detectability or bony detail. At lower resolution ($a_{\text{vox}}=1$ mm; not shown for space considerations), the CNR results of Figs. 6(d) and 7(d) suggest improved performance with grids; however, observers generally noted no appreciable difference between the grid and no-grid cases, with only one observer noting slightly better muscle-fat differentiation in the case of the grid. This may be because the CNR is already so high (e.g., CNR>5) that the imaging task is too conspicuous to properly differentiate between the two configurations, and further improvement in CNR relates very little to improved observer performance. More selective measures are the subject of future investigation, including the response of human observers (e.g., accuracy and reproducibility in target contouring by a radiation therapist) and machine algorithms (e.g., an automated tissue segmentation algorithm).

IV. DISCUSSION AND CONCLUSIONS

The influence of antiscatter x-ray grids on image quality in cone-beam CT was evaluated in a fairly comprehensive experimental investigation across various body sites, scatter conditions, dose, and spatial resolution. Studies involved linear grids in combination with a FPI on a system for kilovoltage cone-beam CT imaging and guidance of radiation therapy. Grids were found to be effective in reducing x-ray scatter “cupping” artifacts, with heavier grids providing increased image uniformity (Figs. 2 and 3). The system was found to be highly robust against ring artifacts that might arise in CT reconstructions as a result of gridline shadows in the projection data. Even under gross perturbations of the grid and correspondingly poor flood-field corrections, axial reconstructions exhibited only slight degradation due to artifact, with ring artifacts approximately at the level of the voxel noise (Fig. 4). We attribute this robustness to a sort of 3D Bucky effected by small imperfections in geometric calibration that smear out the gridline shadows on backprojection. The influence of grids on soft-tissue detectability was evaluated quantitatively in terms of CNR (Figs. 5–8) and qualitatively in patient images (Figs. 9–12). The results suggest that there is little strong motivation for the use of grids in cone-beam CT other than reduction of artifacts. The results highlight the tradeoffs in contrast and noise imparted by grids, showing improved image quality and CNR only under conditions of high x-ray scatter, high imaging dose, and low spatial resolution.

A. A race between contrast and noise

Incorporation of an antiscatter grid typically imparts a tradeoff between improved subject contrast and increased
image noise. While reduction of scatter artifacts is certainly an important issue, the use of a grid is most likely to be advisable under conditions where the improved contrast outpaces the increased noise. This is illustrated in Fig. 13(a), where absolute contrast (adipose solid water) and voxel noise are plotted separately versus GR for the Body phantom. For the higher resolution case \( (a_{\text{vox}} = 0.5 \text{ mm}) \), noise increases more rapidly than contrast, and use of a grid degrades CNR. For the lower resolution case, however \( (a_{\text{vox}} = 1 \text{ mm}) \) the improvement in contrast outpaces the noise penalty, and use of a grid will improve image CNR. Similarly in Fig. 13(b) for the Head phantom, where contrast (breast to solid water) outpaces the noise only in the lower-resolution \( (a_{\text{vox}} = 1.0 \text{ mm}) \) case.

These results indicate that antiscatter grids can degrade or improve image quality in cone-beam CT, depending on the anatomical site, scatter conditions, dose, and spatial resolution. Grids always impart an improvement in subject contrast; however, there is typically a corresponding increase in image noise that may be compensated by either increased imaging dose or reduced spatial resolution in order to obtain an improvement in CNR. The dependence of image noise on dose and voxel size is well understood in transaxial CT \(^{28,29}\) and, more recently, in cone-beam CT.\(^{4,30}\) Since the dependence of image noise on voxel size is stronger than the dependence on dose (viz., inverse cube-root and square-root dependence, respectively),\(^{28}\) the latter is perhaps a more efficient strategy to manage the noise penalty, but only in cases where reduction in spatial resolution is consistent with the imaging task. Overall, these results suggest that grids improve image quality under conditions of high x-ray scatter (i.e., large FOV\(_z\) and/or large anatomical sites), where the detector is input-quantum limited either by means of high imaging dose or reduced spatial resolution. In the majority of cases considered above, including Head and Body sites imaged at low dose (<2 cGy) and sub-millimeter spatial resolution, grids had little effect or were detrimental to image quality, particularly if x-ray scatter magnitude could be managed by reducing FOV\(_z\) to the minimum required to cover the region of interest. These findings were borne out quantitatively in terms of CNR (Figs. 5–7) and qualitatively in patient images (Figs. 9–12).

B. Strategies for x-ray scatter management in cone-beam CT

X-ray scatter is perhaps the largest physical impediment to cone-beam CT image quality, and at least four strategies for scatter management can be brought to bear on the problem: (1) optimization of imaging geometry (air gap),\(^{7}\) (2) limiting FOV\(_z\) (i.e., cone angle) to the minimum required to cover the region of interest;\(^{28}\) image processing using scatter-correction algorithms based on model or experimental representations of the scatter distribution;\(^{13–15,31,32}\) and (4) use of an antiscatter grid.

The first represents perhaps the most important strategy, since cone-beam CT accommodates a large air gap by virtue of the rotational geometry, and (similar to the gaps versus grids argument in chest radiography from the 1980s and 90s)\(^ {9–12,33}\) large air gaps are more efficient than heavy grids in reducing x-ray scatter under high-scatter conditions.\(^ {11}\) The second imparts a simple, pragmatic discipline upon the modality: expose no more than you need to see. As reported previously, SPR reduces in proportion to cone angle, so reducing FOV\(_z\) from 25 to 10 cm in the pelvis imparts a dramatic decrease in SPR. For example, in cone-beam CT guidance of prostate radiation therapy at our institution, the protocol is FOV\(_z\) = 10 cm, sufficient to cover the gland while reducing degradation by scatter (e.g., comparing Figs. 9 and 11). The third (scatter-correction algorithms) is an important and developing area for cone-beam CT, drawing on Monte Carlo,\(^ {14}\) analytical models,\(^ {15}\) and experimental estimates\(^ {32}\) of the x-ray scatter distribution from view-to-view and offering significant reduction of scatter artifacts. Our experience is that almost any scatter correction can reduce scatter effects significantly—e.g., estimation and subtraction of scatter as a constant or gaussian distribution from each projection, with magnitude determined from a simple lookup of SPRs.\(^ {6}\) More sophisticated, fast, and practical algorithms are anticipated in future work. The last strategy (antiscatter grids) offers improvement in scatter artifacts (cupping) and contrast, but appears to provide improved CNR and overall image quality.
only under high-dose/low-resolution conditions and is detrimental to image quality in most cases considered.

While grids in combination with FPIs are in common use in radiography, their role in cone-beam CT has been unclear for a variety of reasons. First among these is dose, which is already high (~1 cGy) in cone-beam CT and would scale in proportion to the Bucky factor if conventional wisdom prevailed. Second, the potential for interference artifacts between gridlines and the detector matrix is of concern, possibly causing ring artifacts in axial reconstructions, but largely disproven in the results above. Neither of these considerations proved to be significantly limiting factors: the transmission factors (0.6–0.7) and the data of Figs. 5–7 suggest a fairly small dose penalty (less than a factor of 2 for all grids investigated) and little reason to increase dose by a Bucky factor, since the FPI was input-quantum limited under most circumstances (i.e., degradation in noise-equivalent quanta imparted by the grid did not significantly degrade detectability). Still, there is very little indication that the use of grids improves imaging performance under the conditions investigated. The only case in which grids demonstrably improved performance was under the condition of high x-ray scatter (large object and FOV,) combined with high imaging dose (4 mAs/projection) and low-resolution reconstruction (1 mm voxels). In that case, the data suggest selection of a heavy grid (e.g., up to GR=12:1). In most cases, however, particularly at lower doses (e.g., higher numbers of projections) and lower scatter conditions (reduced FOV,), the use of a grid was a detriment to image quality.

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