Antiscatter grids in mobile C-arm cone-beam CT: Effect on image quality and dose

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Purpose: X-ray scatter is a major detriment to image quality in cone-beam CT (CBCT). Existing geometries exhibit strong differences in scatter susceptibility with more compact geometries, e.g., dental or musculoskeletal, benefiting from antiscatter grids, whereas in more extended geometries, e.g., IGRT, grid use carries tradeoffs in image quality per unit dose. This work assesses the tradeoffs in dose and image quality for grids applied in the context of low-dose CBCT on a mobile C-arm for image-guided surgery.

Methods: Studies were performed on a mobile C-arm equipped with a flat-panel detector for high-quality CBCT. Antiscatter grids of grid ratio (GR) 6:1–12:1, 40 lp/cm, were tested in “body” surgery, i.e., spine, using protocols for bone and soft-tissue visibility in the thoracic and abdominal spine. Studies focused on grid orientation, CT number accuracy, image noise, and contrast-to-noise ratio (CNR) in quantitative phantoms at constant dose.

Results: There was no effect of grid orientation on possible gridline artifacts, given accurate angle-dependent gain calibration. Incorrect calibration was found to result in gridline shadows in the projection data that imparted high-frequency artifacts in 3D reconstructions. Increasing GR reduced errors in CT number from 31%, thorax, and 37%, abdomen, for gridless operation to 2% and 10%, respectively, with a 12:1 grid, while image noise increased by up to 70%. The CNR of high-contrast objects was largely unaffected by grids, but low-contrast soft-tissues suffered reduction in CNR, 2%–65%, across the investigated GR at constant dose.

Conclusions: While grids improved CT number accuracy, soft-tissue CNR was reduced due to attenuation of primary radiation. CNR could be restored by increasing dose by factors of ~ 1.6 – 2.5 depending on GR, e.g., increase from 4.6 mGy for the thorax and 12.5 mGy for the abdomen without antiscatter grids to approximately 12 mGy and 30 mGy, respectively, with a high-GR grid. However, increasing the dose poses a significant impediment to repeat intraoperative CBCT and can cause the cumulative intraoperative dose to exceed that of a single diagnostic CT scan. This places the mobile C-arm in the category of extended CBCT geometries with sufficient air gap for which the tradeoffs between CNR and dose typically do not favor incorporation of an antiscatter grid. © 2012 American Association of Physicists in Medicine. [DOI: 10.1118/1.3666947]

Key words: mobile C-arm, image-guided surgery, flat-panel detector, cone-beam CT, x-ray scatter, antiscatter grid, spine surgery, minimally invasive surgery, image quality

I. INTRODUCTION

Image-guided surgery based on high-quality cone-beam CT (CBCT) with a mobile C-arm incorporating a flat-panel detector (FPD) has shown promise in head-and-neck interventions and spine surgery, where CBCT could improve surgical precision in minimally invasive procedures, provide up-to-date image data for surgical navigation and reduce the heavy reliance on fluoroscopy.^{1,2} Recent work has focused on establishing the imaging and radiation dose characteristics of such systems, showing that soft-tissue visibility suitable to surgical guidance is achievable.^{1,2} Among the main factors limiting image quality is x-ray scatter, particularly in

larger body sites such as the thorax and abdomen, where the scatter fluence may exceed the primary fluence.³ Scatter has been shown to reduce the contrast of soft-tissue structures in CBCT, increase image noise, and introduce streak and non-uniformity artifacts.^{3,4}

Methods addressing the x-ray scatter challenge include beam blocking arrays, partially attenuating blocking patterns, estimation of scatter fluence from collimator shadows, Monte Carlo techniques, and antiscatter grids.⁵ Previous work demonstrates the potentially positive effect of antiscatter grids in systems involving a small air gap (e.g., dental, maxillofacial, and musculoskeletal scanners),⁶ while systems

with large air gaps (e.g., image-guided radiation therapy) showed little or no overall benefit from grids in terms of contrast-to-noise ratio (CNR) per unit dose and overall image quality.⁴ The geometry of a mobile C-arm resides somewhere in between with respect to source-detector distance (SDD). This paper addresses whether 3D image quality benefits from incorporation of an antiscatter grid in mobile C-arm CBCT. Analysis includes potential artifacts arising from grid orientation, effect on Hounsfield unit (HU) accuracy, image noise, CNR, overall artifact reduction, and radiation dose. Evaluations are performed in the context of body surgery (e.g., thoracic and lumbar spine surgery), where the scatter fraction is high and the “air gap” can be fairly small, even in CBCT geometry with C-arm isocenter placed within the patient.

II. METHODS AND MATERIALS

II.A. Experimental setup

II.A.1. Mobile C-arm

The imaging platform was the prototype mobile isocentric C-arm for intraoperative CBCT shown in Fig. 1 (PowerMobil, Siemens, Erlangen, Germany) equipped with a FPD [PaxScan 3030 + (1536 × 1536 pixels at 194 μm pitch), Varian Imaging Products, Palo Alto, CA] calibrated to account for dark current offset, stationary gain corrections, and angle-dependent gain corrections.^{1,2,7} The C-arm rotates about the OR table covering a total angular range of ~178° and operates in pulsed-fluoroscopic mode at 3.3 p/s and 7 ms pulse width, tube voltage of 80–120 kVp, and 60–920 mAs. Tube current (0.6–9.2 mA) refers to the time-averaged effective current for continuous fluoroscopy, and mAs is the product of the effective current and the scan time. Scan protocols have been established for various body sites and imaging tasks, varying in kVp, mA,

number of projections, and scan time. Image reconstruction with a volume size of 15 × 15 × 15 cm³ (irradiation field at isocenter 15 × 15 cm²) and 0.3 × 0.3 × 0.3 mm³ voxels used a variation of the Feldkamp–Davis–Kress (FDK) algorithm with geometric calibration. Slice averaging was optionally employed to reduce noise. The resulting images allowed soft-tissue visibility with discrimination of fat, muscle, water, tendon, and liver at doses sufficiently low to permit repeat intraoperative scanning under defined protocols and workflows.^{1,8} Four scan protocols (Table I) were identified previously¹ for CBCT-guided spine surgery and were the nominal protocols for the studies of x-ray scatter detailed below. Measurements to determine associated radiation dose were performed using thorax and abdomen phantoms (QRM GmbH, Erlangen, Germany) sandwiched in between CTDI head and body phantoms (Gammex, Middleton, WI) to account for scatter tails outside the FOV. The thorax and abdomen phantoms were modified to allow measurements at the isocenter and at the periphery by inserting through holes (10 mm diameter). The peripheral measurement locations were at 3, 6, 9, and 12 o'clock positions and had a 12.5 mm center-to-surface distance. All measurements were performed with the phantom positioned at isocenter, and the ion chamber (0.6 cc, 10 × 06, Radcal, Monrovia, CA) was positioned at either the isocenter (centered measurements) or with an in-plane offset (peripheral measurements) and connected to a multipurpose meter (9096, Radcal, Monrovia, CA). The total exposure measured per scan (X_{center} , X_{periph}) at the center or periphery was corrected for temperature and pressure factors and converted to air-kerma (K_{center} , K_{periph}) by

$$\{K_{center}, K_{periph}\}(\text{cGy}) = \{X_{center}, X_{periph}\} \cdot f, \quad (1)$$

where $f = 0.873 \text{ cGy/R}$. The resulting air-kerma provided a reasonable estimate of scan dose at these energies. We defined a position-weighted dose index computed in a manner analogous to the conventional $CTDI_w$,

FIG. 1. (a) Experimental setup for antiscatter grids applied to mobile C-arm CBCT. (1) Antiscatter grid mounted on the FPD. (2) Phantom setup (a head-equivalent extension was removed for better visualization). (3) X-ray tube shown at the start-scan position. (b) Thorax phantom; (c) reconstruction of tissue-equivalent inserts in the thorax phantom.

TABLE I. Technique settings and dosimetry for task-specific protocols in the thorax and abdomen. “Bone protocol” refers to a lower dose technique sufficient for high-contrast bone visualization. “Soft-tissue protocol” refers to a higher dose technique sufficient for soft-tissue visualization.

	Bone protocol		Soft-tissue protocol	
	Thorax	Abdomen	Thorax	Abdomen
kVp	100	100	100	120
mAs	60	130	230	460
# Projections	200	200	200	400
$D_w(\text{mGy})$	1.91	3.70	4.61	12.50

$$D_w(\text{cGy}) = \frac{1}{3}K_{\text{center}} + \frac{2}{3}\bar{K}_{\text{periph}}, \quad (2)$$

where $\bar{K}_{\text{periph}}$ is the average of the air-kerma measured at the four peripheral points.^{9,10}

II.A.2. Antiscatter grids

Antiscatter grids were evaluated with grid ratios (GRs) of 6:1, 8:1, 10:1, and 12:1 with 40 lines-per-cm (Pb vanes with Al interspacing, Soyee Products linear Grids, Korea). The grids lead septal thickness was 0.05 mm with 0.2 mm interspacing material (Al). Height of the septal lines was 6:1–1.2 mm, 8:1–1.6 mm, 10:1–2.0 mm, and 12:1–2.4 mm. Grid thickness was ~ 2.5 mm (6:1–10:1) and ~ 3.25 mm (12:1) with transmission factors ranging from 0.59 to 0.46. The transmission of the grids was determined as follows. First, projection image sequences ($n=100$) were acquired with and without antiscatter grids in place [200 mm polymethyl methacrylate (PMMA) in beam path]. Each sequence was corrected for pixel-dependent offset, and an average image was calculated. The transmission factor was then determined as the mean signal ratio between grid to no-grid images. Grid size was $38.1 \times 45.7 \text{ cm}^2$, with each completely covering the FPD. Grids were aligned and centered on the FPD within ~ 2 mm using external lasers. Due to mechanical flex of the C-arm gantry during rotation, the location of the central ray on the FPD changes with acquisition angle and necessitates an angle-dependent gain calibration that simultaneously corrects for variations in transmission owing to misalignment of the grid (FPD) with the central ray. Detector offset and angle-dependent gain calibration were performed separately for each grid, and the grid was not perturbed between calibration and the imaging studies. In keeping with the desire to minimize patient dose in image-guided surgery according to protocols in Table I, a bucky factor was not applied, although the resulting noise and CNR were considered in relation to what increase in dose (inversely proportional to the square root of the noise) would be required to account for blocked primary radiation.

II.A.3. Imaging phantoms

To assess HU accuracy and noise, a uniform water-equivalent insert in a quality assurance phantom (15 cm diameter, Catphan, The Phantom Laboratories, Inc., Salem,

NY) was used. For quantitative analysis of soft-tissue CNR, an oblate abdominal and an oblate thoracic phantom [$20 \times 32 \times 10 \text{ cm}$, both QRM GMBH, Erlangen, Germany, Figs. 1(b) and 1(c)] were used. The removable center of the oblate phantoms (solid water-equivalent material, 10 cm diameter) was replaced with an acrylic cylinder (electron density relative to water, $\rho_e = 1.15$, 130 HU) containing four tissue-equivalent inserts. The inserts were cortical bone [$\rho_e = 1.69$, 1073 HU contrast to cylinder (ctc)], breast ($\rho_e = 0.96$, -171 HU ctc), liver ($\rho_e = 1.07$, -47 HU ctc), and adipose ($\rho_e = 0.93$, -214 HU ctc), each having a diameter of 28 mm (Gammex RMI, Madison, WI).¹¹ The setup was complemented at superior and inferior ends by CTDI head-and-body phantoms to capture scatter tails extending longitudinally outside the field of view. A custom anthropomorphic body phantom was used to examine the effect of grids on artifacts and overall image quality in a semirealistic context. The phantom was based on a Rando phantom (The Phantom Laboratory, Greenwich, NY) modified to include a variety of soft-tissue-equivalent organ inserts (e.g., liver) and interventional tools (e.g., spine screws).

II.B. Image quality evaluation

II.B.1. Grid orientation

As noted by Siewerdsen *et al.*⁴ imperfect gain-offset corrections and imprecise geometric calibration in combination with an antiscatter grid can lead to artifacts in CBCT reconstructions, with the sensitivity to artifacts depending on the orientation of the grid (direction of residual gridlines). To test the robustness of the C-arm to such effects, a 10:1 grid was aligned with gridlines running along the detector u - and v -directions (see Fig. 1)—perpendicular to or parallel to the axis of rotation, respectively. As in all other experiments, separate gain-offset calibrations were performed in each case. CBCT images of the water-equivalent Catphan insert were acquired and assessed in terms of noise, artifact, and uniformity.

II.B.2. Noise and CT number accuracy

X-ray scatter causes underestimation of attenuation coefficient (μ) in CBCT reconstructions. The accuracy was measured with each grid in CBCT images of a homogeneous solid water cylinder by computing the ensemble mean attenuation coefficient $\langle \mu \rangle$ in regions of interest (ROIs, 10×10 voxels) centered in axial slices across 2 cm about the central axial plane (0.3 mm slice thickness, giving 67 slices). The mean attenuation coefficient was compared to a fixed value approximated from a model of the x-ray spectrum¹² and data from the manufacturer ($\mu_{H_2O} = 0.020 \text{ mm}^{-1}$) to yield a figure termed “inaccuracy”

$$\Delta = \frac{\langle \mu_{x:\text{Grid}} \rangle - \mu_{H_2O}}{\mu_{H_2O}} \times 100. \quad (3)$$

Although the truth value is subject to systematic error in the x-ray spectrum model and manufacturer data, we were only concerned with variation in Δ as a function of GR. All images in Δ analysis were acquired at 100 kVp, so

systematic error in μ_{H_2O} is canceled out in the normalization of Eq. (1), with Δ a reasonable, relative indication of % inaccuracy. Image noise was similarly determined by calculation of the standard deviation in the same ROIs across 67 slices (201 mm) and determining the ensemble average $\langle \sigma \rangle$. To assess HU accuracy and noise, the uniform water-equivalent insert in a quality assurance phantom was encased within a surrounding bolus of soft-tissue-simulating material (Super-Flab, Mick Radio-Nuclear Instruments, Mount Vernon, NY) approximating the size of a typical “body” phantom—with thickness about the phantom of 6 cm anterior, 2 cm posterior, and 2 cm lateral on each side.

II.B.3. Contrast-to-noise ratio

For quantitative analysis of soft-tissue CNR, the oblate abdominal phantom and the oblate thoracic phantom were used. In each case, a 10×10 voxel ROI was placed at the center of each insert and in a region of adjacent background (acrylic) at the same distance from the center of reconstruction across 67 slices (201 mm). The mean (μ) and standard deviation (σ) in each ROI was computed, followed by the ensemble average of $\bar{\mu}$ across all slices in the tissue insert and background ($\langle \bar{\mu}_{tissue} \rangle$ and $\langle \bar{\mu}_{bckgrnd} \rangle$, respectively) as well as the ensemble average standard deviation ($\langle \sigma_{tissue} \rangle$ and $\langle \sigma_{bckgrnd} \rangle$, respectively). The CNR was given by the absolute value of the ratio

$$CNR = \frac{|\langle \bar{\mu}_{tissue} \rangle - \langle \bar{\mu}_{bckgrnd} \rangle|}{\frac{1}{2} (\langle \sigma_{tissue} \rangle + \langle \sigma_{bckgrnd} \rangle)}. \quad (4)$$

CNR was analyzed in CBCT images (using $0.3 \times 0.3 \times 0.9$ mm³ voxels) acquired using the scan protocols of Table 1 for all grids.

III. RESULTS

III.A. Grid orientation

Grid orientation did not impart a substantial effect on CBCT image quality. Following proper correction of the projection data by offset calibration, gain calibration, and angle-dependent gain (delta) calibration,⁷ grid lines were effectively removed from the projection data and imparted no perceivable artifacts in 3D reconstructions. The data were similarly robust against gridline artifact in either grid orientation (i.e., aligned along detector rows or columns) and did not influence CNR. All studies below were conducted with the grid oriented along the detector v -direction. High frequency artifacts may occur in 3D reconstructions when grid line shadows are not completely removed from the projections due to incorrect, out-of-date, or omitted stationary and angle-dependent gain calibration (Fig. 2).

III.B. Noise and CT number accuracy

The HU inaccuracy, Δ , improved (decreased) with increasing grid ratio. As shown in Fig. 3, Δ improved from $\sim 37\%$ for the gridless configuration to $\sim 21\%$ for a 6:1 grid and $\sim 12\%$ for the 12:1 grid, while $\langle \sigma \rangle$ increased by $\sim 33\%$ for the 6:1 grid and $\sim 56\%$ for the 12:1 grid.

III.C. Contrast-to-noise ratio

The CNR in the high-contrast bone insert exhibits minimal change with grid ratio and was conspicuous in all cases, attributable to the high attenuation. However, a significant drop in CNR was observed for all of the soft-tissue-equivalent inserts for all scan protocols. As shown in Fig. 4, the potential benefit for contrast is outweighed by the increase in

FIG. 2. Zoomed and magnified projection image showing (a) an example of gridline aliasing stemming from incorrect angle-dependent gain calibration, and (b) the removal of such shadows/aliasing with correctly applied angle-dependent gain calibration. Grid line artifacts in projection images cause high frequency artifacts in the reconstructions (c), an effect that does not occur in correctly processed images (d).

FIG. 3. Hounsfield unit inaccuracy, Δ (left axis), and voxel noise (right axis) as a function of grid ratio. Grids are seen to improve HU accuracy but impart an increase in voxel noise unless accompanied by an increase in exposure technique.

voxel noise, and grids imparted a monotonic reduction in CNR as a function of grid ratio. For the soft-tissue protocol in the thorax, the CNR of adipose, breast, and liver reduced by 12.4%, 32.2%, and 65.9%, respectively, from the gridless configuration to the 12:1 grid. Similarly for the soft-tissue

protocol in the abdomen, the CNR of adipose, breast, and liver reduced by 13.2%, 28.8%, and 65.9%, respectively, from the gridless configuration to the 12:1 grid. Analysis in CBCT images reconstructed at finer or coarser voxel size did not change the result. The results are illustrated qualitatively in the images of Fig. 5, cropped about the region of the tissue-equivalent insert and equivalently window-leveled for intercomparison. Obvious is the reduction in shading artifacts with increased grid ratio. In the “race” between improved contrast and increased noise, grids appear to impart a net degradation on CNR for the C-arm geometry. Assuming that noise is inversely proportional to the square root of dose, an increase in dose by a factor of ~ 1.63 – 2.23 for the thorax and ~ 1.73 – 2.49 for the abdomen (depending on GR) would be required to restore CNR to that achieved in the gridless configuration.

Figure 6 illustrates similar findings in the context of an anthropomorphic body phantom showing the overall qualitative effect of a grid on artifact, noise, and low-contrast visibility. The axial image shows a slight improvement in overall uniformity (shading artifact) with the grid. However, there is little or no benefit to image quality overall, and an increase in noise in the case of the grid (at equivalent patient dose) is appreciable at a level that could diminish soft-tissue visibility.

FIG. 4. CNR as a function of grid ratio for (a) thorax phantom bone protocol, (b) thorax phantom soft-tissue protocol, (c) abdomen phantom bone protocol, and (d) abdomen phantom soft-tissue protocol. At fixed patient dose (see Table I) grids impart a significant reduction in soft-tissue CNR.

FIG. 5. Tissue-equivalent inserts in the thoracic and abdomen phantoms for the bone and soft-tissue protocols at various grid ratios. The images correspond to the CNR measurements of Fig. 4. A reduction in scatter artifacts extending vertically from the cortical bone insert toward the spine insert is visible. The horizontal high- and low-intensity streak artifacts around the cortical bone insert arise from the incomplete orbit of the C-arm ($\sim 178^\circ$).

IV. DISCUSSION

This work presented a summary analysis of the effect of antiscatter grids on image quality in C-arm CBCT, specifically in the context of orthopaedic (spine) image-guided surgery. The work is complementary to the existing body of literature on evaluation of antiscatter grids in CBCT.^{3–6} Given accurate offset calibration, gain calibration, and angle-dependent gain calibration, the orientation of the grid with respect to the detector (i.e., along the u or v direction) had no impact on 3D image quality and CNR, in agreement with the

findings by Siewerdsen *et al.*^{3,4} However, the applied calibration steps, offset, stationary, and angle-dependent gain, need to be carefully performed to prevent high-frequency artifacts in the reconstruction stemming from uncorrected grid line shadowing and aliasing. Grids demonstrated a monotonic improvement in HU accuracy as a function of grid ratio—for example, reducing the HU inaccuracy from $\sim 30\%$ – 40% for the gridless configuration to $\sim 2\%$ – 20% for the highest grid ratio. A corresponding increase in voxel noise was observed—up to $\sim 50\%$ – 60% . Soft-tissue CNR suffered overall with the introduction of antiscatter grids, demonstrating a monotonic reduction in CNR as a function of grid ratio: $\text{CNR}_{\text{adipose}}$ reduced by 2%–11% (thorax) and 3%–20% (abdomen); $\text{CNR}_{\text{breast}}$ and $\text{CNR}_{\text{liver}}$ reduced by up to 65% (thorax) and 53% (abdomen) depending on grid ratio. Qualitative inspection of image quality in phantoms confirmed an increase in noise that outweighed the potential benefits for contrast. The field size was fixed at $15 \times 15 \text{ cm}^2$ throughout, corresponding to the $30 \times 30 \text{ cm}^2$ detector used for CBCT. Larger detectors (e.g., $40 \times 30 \text{ cm}^2$) would entail a larger field size and correspondingly higher scatter magnitude. The results overall suggest a detrimental effect of antiscatter grids on 3D image quality, thus suggesting that the grid would ideally be removed (preferable automatically due to sterile draping of the C-arm during surgery) when switching from fluoroscopic into CBCT mode.

V. CONCLUSION

The results demonstrate that while antiscatter grids improve uniformity and HU accuracy in CBCT reconstructions, these are not the critical metrics of performance in image-guided surgery. Rather, soft-tissue visibility at a given level of radiation dose is of paramount importance, and grids were shown to offer little overall benefit in this regard. An increase in dose by factors of ~ 1.6 – 2.5 (depending on GR) would be required to restore soft-tissue CNR to the level achieved in the gridless configuration. This is an unattractive solution in the context of repeat intraoperative scanning, where minimum-dose protocols for soft-tissue visualization (e.g., no grid: 4.6 mGy thorax, 12.5

FIG. 6. Axial and sagittal views of the anthropomorphic body phantom with implanted spine hardware (resolution $0.3 \times 0.3 \times 0.9 \text{ mm}^3$). Images with a 10:1 grid in place (c, d) exhibit little or no improvement in overall image quality compared to the gridless images (a, b) and show a slight increase in image noise unless the dose to the patient is increased to counter the loss in primary fluence.

mGy abdomen) are important to clinical acceptance, and the marginal benefits to artifacts and HU accuracy do not likely warrant such a dose increase in image-guided surgery. Any number of scatter correction techniques, ranging from simple parametric estimates of a low-frequency scatter fluence to more sophisticated beam blocker pattern, and Monte Carlo scatter correction techniques are anticipated to perform at least as well as grids with respect to image uniformity and CNR.⁵ With respect to previous work examining the performance of grids in other CBCT applications—e.g., compact dental or musculoskeletal scanners and long-geometry IGRT scanners—these results place the C-arm in the latter category of “good geometry” characterized by a large source-detector distance, high magnification, and large air gap, which seminal work by Neitzel¹³ on scatter modeling and adaptation of such models to CBCT by Siewerdsen and Jaffray¹⁴ demonstrate as the first, best defense against scatter. The results suggest little or no benefit—in fact, a detriment—to 3D image quality by incorporation of antiscatter grids in C-arm CBCT. This is contraindicated to the current common practice of using antiscatter grids in II and FPD based fixed and mobile C-arms commonly deployed for fluoroscopy. For systems intended to provide both fluoroscopy and CBCT, with the former mode employing a grid, this raises a logistical question of a grid that could be in place for fluoroscopy but removed for 3D imaging. Considering the sensitivity of flood-field corrections to grid placement, such an arrangement could introduce further degradation to image quality.

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