Two-dimensional antiscatter grid: A novel scatter rejection device for Cone-beam computed tomography

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Purpose: Scattered radiation remains to be a major cause of image quality degradation in Flat Panel Detector (FPD)-based Cone-beam computed tomography (CBCT). We have been investigating a novel two-dimensional antiscatter grid (2D-ASG) concept to reduce scatter intensity, and hence improve CBCT image quality. We present the first CBCT imaging experiments performed with the 2D-ASG prototype, and demonstrate its efficacy in improving CBCT image quality.

Methods: A 2D-ASG prototype with septa focused to x-ray source was additively manufactured from tungsten and mounted on a Varian TrueBeam CBCT system. CBCT projections of phantoms were acquired with an offset detector geometry using TrueBeam's "developer" mode. To minimize the effect of gantry flex, projections were gain corrected on angle-specific bases. CBCT images were reconstructed using a filtered backprojection algorithm and image quality improvement was quantified by measuring contrast-to-noise ratio (CNR) and CT number accuracy in images acquired with no antiscatter grid (NO-ASG), conventional one dimensional antiscatter grid (1D-ASG), and the 2D-ASG prototype.

Results: A significant improvement in contrast resolution was achieved using our 2D-ASG prototype compared to results of 1D-ASG and NO-ASG acquisitions. Compared to NO-ASG and 1D-ASG experiments, the CNR of material inserts improved by as much as 86% and 54% respectively. Using 2D-ASG, CT number underestimation in water equivalent material section of the phantom was reduced by up to 325 HU when compared to NO-ASG and up to 179 HU when compared to 1D-ASG.

Conclusion: We successfully performed the first CBCT imaging experiments with a 2D-ASG prototype. 2D-ASG provided significantly higher CT number accuracy, higher CNR, and diminished scatter-induced image artifacts in qualitative evaluations. We strongly believe that utilization of a 2D-ASG may potentially lead to better soft tissue visualization in CBCT and may enable novel clinical applications that require high CT number accuracy. © 2017 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.12724]

Key words: anti-scatter, CBCT, grid, two-dimensional

1. INTRODUCTION

Scattered x-rays in CBCT reduce soft tissue visualization, CT number accuracy, and introduce nonuniformity artifacts. While current strategies for x-ray scatter management, such as software-based scatter corrections and radiographic antiscatter grids (1D-ASG)^{1–4} help to reduce the detrimental effects of scatter, desired CT number accuracy, and low-contrast resolution have not been achieved.

Recently, we have developed a novel 2D-ASG prototype for flat panel detector (FPD)-based CBCT, and demonstrated a reduction in x-ray scatter intensity. We demonstrated an improvement in primary transmission characteristics in projections acquired with the 2D-ASG compared with projections acquired with 1D-ASG. With respect to 1D-ASG, projections acquired with 2D-ASG displayed a factor of 3–6 lower scatter to primary ratio, and a factor of 1.3–1.4 improvement in CNR.⁵ In this letter, we present the first CBCT images produced from projections acquired using the 2D-ASG prototype, and report on the improvement in CT

number accuracy, reduction of image artifacts, and improvement in CNR.

2. MATERIALS AND METHODS

2.A. Two-dimensional antiscatter grid design and setup

The 2D-ASG prototype was fabricated from Tungsten by Philips (Best, Netherlands) using a unique Powder Bed Laser Melting (PBLM) additive manufacturing process. Two 2×20 cm² 2D-ASG modules were individually manufactured and joined to make a single 2×40 cm² 2D-ASG prototype displayed in Fig. 1(a). The 2D-ASG prototype has rectangular holes, septal height of 23 mm, septal thickness of 0.1 mm, grid pitch of 2.91 mm, and grid ratio of 8.2. 2D-ASG septa were aligned to the x-ray source focal point in Varian TrueBeam half-fan CBCT geometry (Varian Medical Systems, Palo Alto, CA, USA). 2D-ASG was first attached to an Al support frame using an adhesive, and then mounted

directly on the PaxScan 4030CB FPD surface in place of the conventional 1D-ASG. Figures 1(a) and 1(b) show the 2D-ASG prototype, which was mounted on an aluminum frame and used in our experiments.

2.B. Experimental setup

The CBCT experiments were performed using a Varian TrueBeam STx linac system with a PaxScan 4030CB FPD (Varex Imaging, Salt Lake City, UT, USA) and a GS-1542 x-ray tube. Varian TrueBeam STx linac system has a default radiographic 1D-ASG in the CBCT system, with a grid ratio of 10, septal thickness of 0.036 mm, and line density of 60 lines/cm. X-ray source to isocenter and source to detector distances were 100 and 52 cm, respectively. To allow for large field of view (FOV) scans, FPD was offset by 16 cm in the transverse direction. The full FPD FOV was irradiated to mimic a clinical scan, and the central $2 \times 40 \text{ cm}^2$ area of FPD with NO-ASG, 1D-ASG, or 2D-ASG was exposed to x rays. The rest of the FPD was shielded by a 1.6 mm thick lead sheet. CBCT projections were acquired from 360° rotation scans, with an interval of roughly 1° between projections, and reconstructed using Feldkamp-Davis-Kress (FDK) algorithm⁶. In order to account for the offset detector geometry during reconstruction, a weighting method described by Cho⁷ and Wang⁸ was implemented along with the FDK algorithm. Alignment of the FPD was verified by acquiring projections at various lateral, longitudinal, and vertical positions prior to performing CBCT experiments. Detector position offset of 5 mm in the vertical, lateral, and longitudinal directions from the expected detector geometry caused about a 1.2%, 5.2%, and 1.9% reduction in the primary transmission respectively when compared to the zero offset position. For this reason, a zero offset FPD position was selected for all of the CBCT experiments as it provided the greatest primary transmission

FIG. 1. (a) 2D-ASG mounted on an aluminum frame (b) Up-close view of the 2D-ASG section marked by a dashed box in (a). [Color figure can be viewed at wileyonlinelibrary.com].

and the least amount of grid septa shadow. To evaluate the reduction in x-ray scatter due to the 2D-ASG alone, software scatter correction methods were not applied in any of the experiments.

Experiments were performed using Catphan 504 phantom (small phantom), Catphan 504 phantom with an annular uniformity sleeve (large phantom), and a pelvis phantom (The Phantom Laboratory, Salem, NY, USA) to mimic patient shape and geometry. Projections were acquired at 125 kVp, 38 mA, 13 ms x-ray tube settings, and with a 0.9 mm Titanium filter. A bow tie filter was not used in any of the experiments to identify the improvements in image quality due to 2D-ASG alone. All of the experiments were performed in Varian “developer” mode using XML scripts. Projections were acquired in “dynamic gain” mode and were normalized using the x-ray tube monitor chamber to minimize the effect of tube output variations. Three sets of CBCT acquisitions were performed for each phantom setup: NO-ASG, native 1D-ASG, and our novel 2D-ASG prototype.

In order to suppress grid septal shadows, a gain correction map was developed from single angle NO-ASG flood projection and both, 1D and 2D-ASG full gantry rotation flood CBCT scans without a phantom:

$$\text{GainMap}(x, y, \beta) = \frac{\text{Flood}_{\text{NO-ASG}}(x, y, 0^\circ)}{\text{Flood}_{\text{ASG}}(x, y, \beta)} \quad 0 < \beta \leq 360 \quad (1)$$

where β is the gantry angle of the projections obtained over the full gantry arm rotation range, $\text{Flood}_{\text{NO-ASG}}(x, y, 0^\circ)$ is the pixel value at each location (x, y) along the FPD in flood NO-ASG projections at a 0° angle, and $\text{Flood}_{\text{ASG}}(x, y, \beta)$ is the pixel value at an angle β and the corresponding location (x, y) in flood projections acquired with 1D or 2D-ASG. Gain maps for 1D-ASG and 2D-ASG were acquired without a phantom and only once prior to all of the experiments. To suppress grid line artifacts (GLAs) in phantom scans, an angle specific gain map in Eq. 1 was applied to phantom CBCT projections by matching the phantom projection gantry angle with the gain map gantry angle:

$$P(x, y, \beta) = P_0(x, y, \beta) * \text{GainMap}(x, y, \beta) \quad (2)$$

where P_0 are the uncorrected phantom projections and P are the phantom projections after angle specific gain map correction.

2.C. Image quality evaluation metrics

The CBCT images were reconstructed with a 46 cm field of view and 1 mm^3 voxel size. Qualitative assessment of reconstructions was performed by visually evaluating the reconstruction of the large phantom material insert sections with 2D-ASG, 1D-ASG, and with NO-ASG. Quantitative assessment was performed by calculating CNR of material inserts and CT number accuracy in the phantom uniformity material section ROIs displayed with dashed circles in Figs. 2(a) and 2(b). CT number mean and variation values

FIG. 2. (a) ROIs in the small phantom. (b) ROIs in the large phantom. (c) ROIs in the pelvis phantom. [Color figure can be viewed at wileyonlinelibrary.com]

were calculated in bone equivalent material in the pelvis phantom ROI displayed with a dashed circle in Fig. 2(c). CNR was calculated in the following way:

$$CNR = \frac{|C - B|}{\frac{1}{2} \sqrt{\sigma_C^2 + \sigma_B^2}} \quad (3)$$

where C is the mean pixel value in the material insert, B is the mean pixel value in the surrounding background ROI, σ_C is the standard deviation of pixel values in the material insert, and σ_B is the standard deviation of pixel values in the surrounding background ROI.

Improvement in CNR was assessed as the difference between CNR measured with the 2D-ASG prototype and with either NO-ASG (2D/NO) or 1D-ASG (2D/1D):

$$\%CNR \text{ Change} = \frac{CNR_{2D-ASG} - CNR_{ASG}}{CNR_{ASG}} \times 100 \quad (4)$$

where CNR_{ASG} corresponds to either CNR_{NO-ASG} or CNR_{1D-ASG} .

Attenuation values were normalized by a water attenuation coefficient that produced 0 HU in a small ROI in the water equivalent section in the 2D-ASG images of the small phantom, and HU value uniformity for the small and the large phantoms were calculated in reference to this value. Mean ROI HU number was calculated using Eq. 5:

$$\bar{I}_{ROI} = \frac{1}{\Omega_{xy}} \sum_y \sum_x I(x, y, z); \quad z = \text{const}, \{z|z \in \Omega\} \quad (5)$$

where, $I(x, y, z)$ is the HU value in a ROI centered at (x, y, z) , Ω_{xy} is the area of the corresponding ROI in the x-y slice, and Ω is the reconstruction volume.

3. RESULTS

First row in Fig. 3 shows the material inserts section in the small phantom images acquired with NO-ASG, 1D-ASG, and 2D-ASG. A cupping artifact is visible in the images

FIG. 3. Top row: small phantom; Middle row: large phantom; Bottom row: pelvis phantom; First column: images acquired with NO-ASG; Second column: images acquired with 1D-ASG; Third column: images acquired with 2D-ASG.

FIG. 4. Large phantom HU profile along the dotted line in Fig. 2(b). [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 5. Pelvis phantom HU profile along the dotted line in Fig. 2(c). [Color figure can be viewed at wileyonlinelibrary.com]

acquired with NO-ASG and 1D-ASG, and is not visible in the images acquired with 2D-ASG. It was determined that the ring artifacts visible in the outer uniformity section in the 2D-ASG reconstruction of the small phantom are due to flickering pixels caused by internal gain calibration in Varian TrueBeam CBCT system, and were not caused by the 2D-ASG.

Second row in Fig. 3 shows images of the material insert section in the large phantom acquired with NO-ASG, 1D-ASG, and 2D-ASG. Again, a heavy cupping artifact is visible in the images acquired with NO-ASG and 1D-ASG, and is not visible in the images acquired with the 2D-ASG. The significant reduction in the cupping artifact, when using 2D-ASG, can also be seen in Fig. 4, which shows the HU value variation in the large phantom along the red dashed line indicated in Fig. 2(b).

A considerable reduction in the cupping artifact can also be seen in the pelvis phantom images acquired with the 2D-ASG compared to the pelvis phantom images acquired with NO-ASG and 1D-ASG. Third row in Fig. 3 displays the

pelvis phantom images acquired with NO-ASG, 1D-ASG, and 2D-ASG. Figure 5 shows profile HU values measured in all three pelvis phantom experiments along the location marked with a red dashed line in Fig. 2(c). In all cases, the contrast of the bone equivalent material was drastically improved with the use of 2D-ASG. The mean HU measure in the bony region improved by 656 ± 72.7 HU when comparing the 2D-ASG pelvis reconstruction to NO-ASG pelvis reconstruction, and 459 ± 64.5 HU when comparing the 2D-ASG pelvis reconstruction to 1D-ASG pelvis reconstruction.

Finally, the HU underestimation in the uniform material section of the small phantom was reduced from the range of -114.4 ± 14.1 HU to -28 ± 5.1 HU in the NO-ASG reconstruction to the range of -38.8 ± 9.9 HU to -0.7 ± 8.4 HU in the 1D-ASG reconstruction, and a range of -6.5 ± 18.5 HU to 8.6 ± 6.9 HU in the 2D-ASG reconstruction. The HU underestimation in the uniform material section of the large phantom was reduced from the range of -397.1 ± 10.3 HU to -43.7 ± 9.2 HU in the NO-ASG reconstruction to the range of -250.4 ± 12.7 HU to -4.3 ± 12.0 HU in the 1D-ASG reconstruction, and a range of -71.9 ± 19.8 HU to -7.1 ± 20.2 HU in the 2D-ASG reconstruction.

Table I lists the CNR values for Teflon, Delrin, and Polystyrene inserts in the small phantom images acquired with NO-ASG, 1D-ASG, and 2D-ASG. Locations of these inserts are indicated in Fig. 2(a). In the small phantom, 2D-ASG provided 35%–54% and 8%–26% improvement in CNR with respect to NO-ASG and 1D-ASG configurations, respectively.

Table II lists the CNR values for Teflon, Delrin, and Polystyrene inserts in the large phantom images acquired with NO-ASG, 1D-ASG, and 2D-ASG. Locations of these inserts are indicated in Fig. 2(b). In the large phantom, 2D-ASG provided 56%–85% and 35%–51% improvement in CNR with respect to NO-ASG and 1D-ASG configurations, respectively.

4. DISCUSSION AND CONCLUSION

We developed and presented the first implementation of a 2D-ASG in FPD-based CBCT. Images acquired with 2D-ASG showed considerable improvements in CNR, HU accuracy, and reduction of image artifacts. While in this study we did not use a software-based scatter correction method, using one in conjunction with the 2D-ASG can further improve image quality metrics, such as CNR and CT number accuracy. Although improvement of HU accuracy is expected

TABLE I. NO-ASG, 1D-ASG, and 2D-ASG CNR values in the small phantom.

ROI	CNR _{NO-ASG}	CNR _{1D-ASG}	CNR _{2D-ASG}	% CNR _{2D/NO} Change	% CNR _{2D/1D} Change
a) Teflon	47.3 ± 3.5	57.8 ± 4.5	73.0 ± 0.4	+54.3 %	+26.3 %
b) Delrin	17.4 ± 0.8	23.5 ± 2.7	25.8 ± 1.4	+48.3 %	+9.8 %
c) Polystyrene	9.9 ± 0.6	12.4 ± 0.3	13.4 ± 1.1	+35.4 %	+8.1 %

TABLE II. NO-ASG, 1D-ASG, and 2D-ASG CNR values in the large phantom.

ROI	CNR _{NO-ASG}	CNR _{1D-ASG}	CNR _{2D-ASG}	% CNR _{2D/NO} Change	% CNR _{2D/1D} Change
a) Teflon	16.7 ± 0.1	19.3 ± 2.2	26.2 ± 0.8	+56.9 %	+35.8 %
b) Delrin	3.5 ± 0.03	4.6 ± 0.05	6.5 ± 0.9	+85.7 %	+41.3 %
c) Polystyrene	3.0 ± 0.3	3.1 ± 0.8	4.7 ± 0.6	+56.7 %	+51.6 %

with the 2D-ASG, an important detail to note is the CNR improvement in the CBCT images. CNR improvements between 8% and 52% are achieved with the 2-D ASG over 1-D ASG results. Our previous studies suggested that the improvement in CNR is due to both higher primary transmission and lower scatter transmission.⁵ Other published studies indicated that, CNR may even be degraded by 1-D ASG in low-scatter imaging conditions due to lower primary transmission of 1-D ASG.^{9–12}

While at this point CT number accuracy in MDCT images is better than in our results, we believe that there is large room for improvement with 2D-ASG, which may be achieved by optimizing the grid geometry. We plan to increase the grid ratio in our future prototypes, which is expected to further reduce scatter transmission and improve HU accuracy. Although our first 2D-ASG prototype only covered a 2 cm wide section of the FPD in the axial direction, the versatility of PBLM will allow for larger 2D-ASGs to be fabricated in the future to cover the entire detection surface of the FPD. Due to the precision of PBLM, scatter and primary transmission characteristics of the larger grid is expected to be similar to the small prototype tested, and image quality improvement is expected to stay consistent.

An important issue during our imaging experiments was the effect of gantry flex, or sag, during CBCT acquisitions; the shadow of grid septa in projections shifts slightly as the x-ray source position changes with respect to FPD position during the gantry rotation. To address this problem, we implemented a gantry angle specific gain correction approach. This approach successfully reduced ring artifacts caused by the gantry flex. However, the robustness of “gantry angle-specific gain correction” may be a concern, where a drift in gantry flex characteristics over time may require frequent gain calibrations. While we have not performed such a long-term study, we have acquired tens of CBCT scans over a period of multiple hours during each experiment session, where new calibration data were acquired during each session. In these sessions, we have not observed a degradation in the performance of our gain correction algorithm.

Due to the grid septa in two dimensions, our 2D-ASG approach requires fixed source-detector geometry. On the other hand, conventional ASGs allow for the movement of FPD in the direction parallel to its septa. This flexibility enables CBCT acquisitions to be performed in either centered or offset detector geometry. In the 2D-ASG case, two separate 2D-ASGs are required to be able to use both centered and offset detector CBCT geometry. In our current implementation, only offset detector geometry scans can be acquired. A possible solution would be to engineer a modular ASG

design, where a different 2D-ASG is mounted on the FPD depending on the acquisition geometry. Alternatively, all of the CBCT scans may need to be acquired with a large field of view geometry. Although 2D-ASG applications in fluoroscopy have not been investigated at this point, we believe that 2D-ASG may also improve the low-contrast resolution in fluoroscopy images. However, such systems typically have varying source-to-detector distance, and may deteriorate primary transmission of 2D-ASG. Such geometry variations may also lead to grid artifacts due to variations in septal shadows. To implement a 2D-ASG in systems with combined CBCT and fluoroscopy capability, 2D-ASGs with lower grid ratios may be better suited, as lower grid ratios are expected to be more tolerant to variations in source-to-detector geometry. Additionally, robust grid artifact correction methods may be required that can address the variations in septal shadows due to varying source-to-detector geometry.

We strongly believe that the implementation of 2D-ASGs in clinical systems can potentially translate to better clinical utilization of CBCT images. In the context of image guided radiation therapy, improvements in soft tissue visualization may lead to better localization of soft tissue targets during radiation delivery. Moreover, high HU accuracy and soft tissue visualization may also enable efficient treatment monitoring and treatment adaptation approaches. High accuracy of CT numbers may enable CBCT-based treatment dose calculations. Improved coherence of CT numbers between CBCT and CT images may also improve the robustness of deformable image registration algorithms, which is key to automated monitoring of tumor and normal tissue changes throughout the course of radiation treatments.

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