Concurrent kilovoltage CBCT imaging and megavoltage beam delivery: suppression of cross-scatter with 2D antiscatter grids and grid-based scatter sampling

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Abstract

Objective. The concept of using kilovoltage (kV) and megavoltage (MV) beams concurrently has potential applications in cone beam computed tomography (CBCT) guided radiation therapy, such as single breath hold scans, metal artifact reduction, and simultaneous imaging during MV treatment delivery. However, MV cross-scatter generated during MV beam delivery degrades CBCT image quality. To address this, a 2D antiscatter grid and a cross-scatter correction method were investigated in the context of high dose MV treatment delivery.

Approach. A 3D printed, tungsten 2D antiscatter grid prototype was utilized in kV CBCT scans to reduce MV cross-scatter fluence during concurrent MV beam delivery. Remaining cross-scatter in projections was corrected by using the 2D grid itself as a cross-scatter intensity sampling device, referred to as grid-based scatter sampling (GSS). To test this approach, kV CBCT acquisitions were performed while delivering 6 and 10 MV beams, mimicking high dose rate treatment delivery scenarios. kV and MV beam deliveries were not synchronized to eliminate MV beam delivery interruption. MV cross-scatter suppression performance of the proposed approach was evaluated in projections and CBCT images of phantoms. Main results. 2D grid reduced the intensity of MV cross-scatter in kV projections by a factor of 3 on the average, when compared to conventional antiscatter grid. Remaining cross scatter as measured by the GSS method was within 7% of measured reference intensity values, and subsequently corrected. CBCT image quality was improved substantially during concurrent kV–MV beam delivery. Median Hounsfield Unit (HU) inaccuracy was up to 182 HU without our methods, and it was reduced to a median 6.5 HU with our 2D grid and scatter correction approach. Our methods provided a factor of 2–6 improvement in contrast-to-noise ratio. Significance. This investigation demonstrates the utility of 2D antiscatter grids and grid-based scatter sampling in suppressing MV cross-scatter. Our approach successfully minimized the effects of MV cross-scatter in concurrent kV CBCT imaging and high dose MV treatment delivery scenarios. Hence, robust MV cross-scatter suppression is potentially feasible without MV beam delivery interruption or compromising kV image acquisition rate.

1. Introduction

In image-guided radiation therapy (IGRT), 2D or 3D kV image guidance has been temporally sequenced with respect to MV treatment beam, such that targets are localized either before or intermittently during treatment delivery (Goyal and Kataria 2014, Korreman 2015). The concept of using kV and MV beams concurrently has potential advantages over temporally sequenced kV and MV beam delivery, and variety of novel applications have been proposed in recent years.

One of the promising applications of concurrent kV–MV beam delivery is in breath-hold cone beam computed tomography (CBCT), where MV and kV projections are acquired simultaneously to reduce CBCT
scan trajectory and duration by half. It has been shown that such kV–MV imaging scheme could reduce scan time down to a mere 15 s which could enable single breath-hold CBCT (Blessing et al 2010, Wertz et al 2010). Furthermore, concurrent kV–MV beam delivery could be used for simultaneous 3D imaging during MV treatment delivery to improve intrafraction target localization and to calculate delivered dose (Ling et al 2011, Iramina et al 2020b, Iramina et al 2021, Boylan et al 2012, Iramina et al 2020a). Likewise, tomosynthesis–like fast 3D imaging approaches have been investigated by acquiring limited-angle kV and MV projections concurrently (Ren et al 2014). Another application of concurrent kV–MV projection acquisition is in metal artifact reduction, which exploits relative immunity of MV projections to metal-induced photon starvation artifacts (Wu et al 2014, Li et al 2013, Altunbas et al 2015).

However, concurrent kV–MV beam delivery results in MV cross-scattered x-rays impinging on the kV flat panel detector (FPD) which deteriorates the image quality. Several solutions have been proposed to address this problem. One suggests triggering kV and MV beam pulses in an alternating sequence, such that MV beam is on while kV beam is on and vice versa (Ling et al 2011). While this approach practically eliminates MV cross-scatter in kV projections, it reduces kV projection acquisition rate by 50% and may increase kV imaging duration. Moreover, such kV–MV pulse synchronization may cause MV beam interruptions at high dose rates typically used in stereotactic body radiation therapy (SBRT). Another approach is to reduce kV beam pulse rate, such that some of the kV projections are acquired while only MV beam is on, and kV beam is off. This method allows measurement of MV cross-scatter in projections acquired without kV beam, and subsequent correction of MV cross-scatter intensity (Van Herk et al 2011). This approach negates MV beam interruptions but reduces kV image acquisition rate and cannot suppress noise due to MV cross-scatter. Model-based MV cross-scatter estimation methods have also been investigated to correct cross-scatter in kV projections (Iramina et al 2020b, Boylan et al 2012). However, these approaches may suffer from the discrepancy in modeled and actual clinical imaging conditions, leading to misestimation of cross-scatter intensity. Furthermore, noise due to cross-scatter is not suppressed in such scatter correction approaches.

In this work, we propose a new, two-stage approach to address MV cross-scatter problem. In the first stage, we implement a 2D antiscatter grid developed for kV CBCT imaging (Alexeev et al 2018, Altunbas et al 2017), to reject MV cross-scattered x-rays. This way, Hounsfield Unit (HU) loss can be partially recovered, and stochastic noise due to cross-scatter can be reduced. In the second stage, we implement a scatter correction method to correct MV cross-scatter that is not stopped by the grid. This method, referred to as grid-based scatter sampling (GSS) (Altunbas et al 2021), Yu et al 2020, utilizes 2D grid as a scatter measurement device, and corrects both kV scatter and MV cross-scatter simultaneously in projections. GSS method can further reduce HU loss, thereby improving HU accuracy and reducing image artifacts. We performed experiments to investigate the utility of 2D grid and GSS method in the context of concurrent kV CBCT imaging and MV beam delivery during SBRT–like radiation treatment scenarios.

Scatter suppression methods used in this work, namely 2D antiscatter grids and GSS, were previously investigated in the context of kV scatter suppression (Altunbas et al 2021, Yu et al 2020). However, the utility of these methods has not been investigated in the context of MV cross-scatter suppression. Therefore, the novelty of the present work is in the robust mitigation of MV cross-scatter problem in concurrent kV and MV beam delivery, by using clinically viable kV and MV beam delivery methods. To the best of our knowledge, a similar solution for MV cross-scatter suppression method has not been proposed. Our work, as explained in the following sections, allows MV cross-scatter suppression while delivering kV and MV beams without pulse synchronization. Therefore, concurrent kV imaging and MV beam delivery can be achieved without interrupting the MV beam and without compromising kV image acquisition rate.

Our evaluations below were focused on suppression of MV cross-scatter in concurrent kV CBCT imaging and high dose MV treatment delivery scenarios, such as SBRT. The rationale for kV CBCT imaging during SBRT delivery is to monitor target position in 3D to reduce dosimetric errors due to organ motion during treatment delivery. In principle, our approach can be extended to high temporal resolution kV imaging techniques, such as tomosynthesis (Ren et al 2014) and 2D kV imaging during radiation therapy, and as well as concurrent kV–MV projection acquisition for fast breath-hold CBCT.

2. Methods

2.1. Antiscatter grid properties
The 2D grid prototype used in this work is a focused tungsten 2D antiscatter grid with a grid ratio of 12, grid pitch of 2 mm, and septal thickness of 0.1 mm (Altunbas et al 2019). Focusing geometry of the prototype was designed for the Varian TrueBeam linac (Varian Medical Systems, Palo Alto, CA) Offset detector CBCT geometry. Primary purpose of the 2D grid prototype is to suppress kV scatter during CBCT scans, and therefore, its physical properties (such as septal thickness) were not altered to increase its MV cross-scatter suppression.
performance. The 2D grid was installed directly on the FPD’s face plate, after removing the default (1D) antiscatter grid.

To quantify the benefit of the 2D grid in MV cross-scatter suppression, a second set of imaging experiments were performed with the default 1D antiscatter grid in the TrueBeam system (Altunbas et al 2017). The default 1D grid is a conventional radiographic antiscatter grid with a grid ratio of 10. It has 1D array of lead septa with 0.036 mm septal thickness and 60 septa per cm. Space between lead septa are filled with fiber spacers.

2.2. Experiment setup

kV CBCT scans were performed in offset detector geometry (i.e. half fan geometry) by using the clinical TrueBeam ‘pelvis’ protocol parameters; 900 projections were acquired at 125 kVp, 1080 mAs per scan with bowtie filter and 0.9 mm titanium foil in place. Since mAs and bowtie filter affect the primary signal intensity and hence the relative contribution of MV cross-scatter in projection signal, a subset of imaging experiments was repeated without bowtie filter. When bowtie filter was not used, mAs was reduced to prevent saturation of the detector signal at the phantom-air boundary. Based on our experience, 450 mAs was used to achieve sufficiently high kV image signal in highly attenuating regions of the phantom, while preventing detector saturation in phantom-air boundary.

Experiments were conducted by using head and pelvis sized electron density phantoms, as well as anatomically more realistic thorax and pelvis phantoms. For each phantom and concurrent kV–MV scan protocol, two CBCT scans were performed; one with 2D and the other with the default 1D grid, where kV image acquisition and MV beam delivery parameters were kept identical for both grid types. For each scan protocol, two more CBCT scans were acquired with kV beam only, one with 2D grid and the other with the default 1D grid. Parameters in kV-only CBCT scans were identical to the kV parameters in the concurrent kV-M CBCT scans. These kV-only scans were MV cross-scatter free, and they served as reference when evaluating the performance of antiscatter grids in MV cross-scatter suppression.

kV CBCT imaging during SBRT delivery was emulated by setting the MV field size to $3 \times 3$ cm$^2$ and delivering 1200 Monitor Units (MU) at a rate of 1200 MU min$^{-1}$ during kV CBCT acquisition, corresponding to 1.33 MU per kV projection. Based on our experience, most targets treated with SBRT have field aperture sizes that are often smaller than $3 \times 3$ cm$^2$ due to MV beam modulation by the multi leaf collimator (MLC). Therefore, our choice of $3 \times 3$ cm$^2$ field size is a conservative approximation to MLC aperture sizes used in SBRT. Depending on the treatment plan, an SBRT treatment beam may contain several hundred to several thousand MUs. Our choice of 1200 MU per treatment beam aims to represent the average treatment beam MU. Likewise, MU rate in a SBRT treatment beam can vary from several hundred MU min$^{-1}$ up to 1400–2400 MU min$^{-1}$, for 6 MV and 10 MV flattening filter free beams, respectively, depending on the treatment plan and delivery technique. Thus, 1200 MU min$^{-1}$ was selected to emulate the average MU rate in SBRT treatments. To evaluate the performance of our methods at very high cross-scatter intensity conditions, the imaging protocol was repeated by increasing the field size to $10 \times 10$ cm$^2$. The effect of beam energy was evaluated by using 6 and 10 MV flattening filter free (FFF) beams, because FFF beams are commonly used in SBRT treatments. In addition, the energy difference between non-FFF and FFF beams is small (Glide-Hurst et al 2013), and therefore, non-FFF beams were not employed in this study.

2.3. Projection postprocessing and image reconstruction

Although concurrent kV projection acquisition and MV beam delivery can be performed in the treatment mode of TrueBeam, concurrent CBCT scan and MV delivery could not be performed (Iramina et al 2021). To address this limitation, TrueBeam’s ‘Developer’ mode was used, where customized, concurrent kV acquisition and MV beam delivery protocols were generated by using XML scripts. kV projections were acquired using ‘DynamicGainFluoro’ mode. In Developer mode, dark field and flat field corrections were applied by the TrueBeam system, however this mode does not allow other post-processing steps, such as scatter correction, beam hardening correction, or image reconstruction. Thus, TrueBeam’s standard data correction methods for clinical CBCT scans, such as scatter correction, were not used in this study.

Subsequent post-processing and image reconstruction were done, after exporting the dark and flat field-corrected projections from TrueBeam. These exported projections are referred to as ‘raw’ projections in the rest of the text. After exporting the projections from the TrueBeam, a gantry angle-specific gain map correction was applied to each projection to reduce the 2D grid shadows in projections. This process was elaborated in a prior publication (Alexeev et al 2018) and section 2.4 below. Briefly, gain map correction is a flat field correction scheme, where 900 flood field projections (projections without phantom, but with antiscatter grid) were acquired while the gantry was rotating as in a CBCT scan. Flood projections characterize the signal intensity variations due to grid’s septal shadows and detector sensitivity nonuniformities. Subsequently, gain correction maps were generated from each flood projection and applied to each raw phantom projection on a gantry
angle-specific basis. For a given CBCT scan protocol, two flood projection sets were acquired, one with 2D and the other with the default 1D grid by using the kV beam only, from whom two gain map sets were generated. While this step is particularly important to reduce septal shadows in projections acquired with the 2D grid (figure 2), the same gain correction step was also applied to 1D grid projections, to assure that image processing steps are identical for both grid types.

Detector pixel size during acquisitions was $0.388 \times 0.388$ mm$^2$. After gain map correction, $3 \times 3$ binning was applied, and CBCT images were reconstructed using filtered backprojection in TIGRE image reconstruction toolkit (Biguri et al 2016). Authors also implemented the offset detector weights scheme, to enable large field of view reconstruction in TrueBeam’s offset detector geometry (Wang 2002). All images were reconstructed using $0.9 \times 0.9 \times 1$ mm$^3$ voxel size. To make sure that image processing steps were identical for 1D and 2D grids, steps described above were applied to all data sets regardless of the grid type.

In addition to 1D grid only and 2D grid only CBCT images, a third set of images were reconstructed by processing 2D grid CBCT projections with the GSS method. These images were referred to as ‘2D grid + GSS’ in the rest of the text, and improvement in image quality by GSS was evaluated with respect to 1D and 2D grid only images. As described in the next section, GSS method corrected the remaining kV and MV cross-scatter in projections acquired with the 2D grid.

2.4. GSS method

Our method employs 2D antiscatter grid as a residual scatter measurement device, which measures and corrects the intensity of residual scatter in projections (Altunbas et al 2021, Yu et al 2020). A brief explanation of the method follows.

When a 2D grid is placed on the detector, pixels located underneath the grid septa receive less x-ray fluence than the pixels located within grid holes (figure 2(a)). Ideally, this effect could be modeled and corrected by getting a second set of projections without the object (referred to as flood projections) to correct such a signal reduction due to 2D grid’s septal shadows in projections (figure 1(b)).

This correction is referred to as gain map (GM), and formulated by

$$GM(x, y) = \frac{C}{F(x, y)},$$

where $C$ is a normalization constant, and $F(x, y)$ is the flood projection. In this work, $C$ corresponds to the mean value of the flood projection set. Thus, septal shadows in a raw projection can be compensated by multiplication with gain maps. However, when residual scatter is present in a CBCT scan, pixels within the grid shadows show a higher intensity than the pixels in a grid hole after gain map correction (figure 2(c)). This effect is due to the additive residual scatter signal in CBCT projections, whereas gain map is a multiplicative correction that is generated from scatter-free flood projections.

In the GSS method, such a signal difference between grid shadows and grid holes is exploited to measure the residual scatter intensity. When residual scatter, $S$, is present in projections, the hyperintense signal intensity pattern in figure 2(c) changes as a function of $S$. Assuming $S$ is piecewise uniform in pixels residing both in grid shadows and grid holes in a small neighborhood of pixels (small neighborhood is a $7 \times 7$ pixel region, corresponding to an area of $2.7 \times 2.7$ mm$^2$), $S$ can be calculated as (Altunbas et al 2021, Yu et al 2020)
Figure 2. (a) A raw phantom projection acquired with 2D grid in place, where grid septal shadows and grid holes are visible. (b) Corresponding flood projection that is used for generating the gain map. (c) Phantom projection after gain map correction. Hyperintense lines at grid septal shadow locations are due to residual kV scatter and MV cross-scatter transmitted through the 2D grid. (d) Residual scatter map is calculated by measuring the signal difference between the hyperintense lines and grid holes in (c). (e) Scatter map is subtracted from the raw phantom projection and gain map is applied. Due to correction of scatter, hyperintense lines are also suppressed in phantom projections.

\[ S(x_i, y_j) = \frac{d(x_i, y_j)}{GM_{grid}(x_i, y_j) - GM_{hole}(x_i, y_j)}, \]

where \( x_i \) and \( y_j \) are for pixels in grid shadows and \( x_s \) and \( y_s \) are for pixels in grid holes, \( d \) is the signal difference in grid shadows and holes in a small neighborhood, \( GM_{grid} \) and \( GM_{hole} \) are the values of gain maps in grid septal shadows and holes, respectively. Based on our empirical evaluations in our previous work, scatter was assumed to be piecewise uniform in a 7 × 7 pixel region, corresponding to 2.7 × 2.7 mm² neighborhood on the detector. Mean scatter intensities calculated with smaller pixel regions (e.g. 3 × 3) were not substantially different, but increased the noise, or standard deviation, in scatter estimations. Therefore, an empirically selected region of 7 × 7 pixels helps to reduce noise in scatter estimations, while estimating residual scatter.

Thus, the two major assumptions in the GSS method are (1) primary kV signal intensity is reduced by 2D grid’s grid shadows, (2) scatter intensity is the same, or uniform, in pixels residing in grid shadows and grid holes, in a small neighborhood of pixels. Previously, these assumptions in the GSS method were shown to be acceptable in suppressing kV scatter (Altunbas et al 2021, Yu et al 2020). These assumptions are still considered valid when MV cross-scatter is present.

Using the above formulations, we could estimate scatter in grid shadows and use interpolation to find residual scatter values in each detector pixel. Finally, calculated scatter (figure 2(d)) was subtracted from projections to achieve scatter corrected projections (figure 2(e)) and proceeded to image reconstruction.

It is important to note that GSS method is fundamentally different than widely adopted beam-stop methods (Altunbas 2014) in the following ways: (1) in beam-stop based methods, beam stops cast a large shadow on multiple pixels, and signal in the shadow is equal to the scatter intensity. Whereas, in our method, the grid wall thickness is only a fraction of a detector pixel size, and the signal in grid shadows is a mixture of both primary and scatter signals. As a result, scatter intensity in 2D grid’s grid shadow cannot be measured directly, as in a beam-stop method. In the GSS method, the scatter intensity was measured via the change in the signal intensity introduced by the 2D grid’s shadows. (2) In beam-stop methods, a beam-stop array is placed between the patient and the x-ray source. In our case, the 2D grid, i.e. the scatter measurement device, is placed between the patient and the detector.

2.5. Measures of comparison
2.5.1. Projection domain evaluations
MV cross-scatter rejection and residual cross-scatter estimation performance were evaluated in projections.

To achieve this, first MV cross-scatter intensity in kV projections was measured during concurrent kV–MV beam delivery, which served as baseline, or reference, to assess the performance of our methods. For a given phantom configuration, two CBCT data sets were acquired, one with kV only and the other with concurrent kV–MV beams. Projections in these two data sets were paired by matching the source angle in each pair. Subsequently, subtraction of kV only projection from kV–MV projection in each pair yielded the reference MV cross-scatter intensity as a function of gantry angle. At each gantry angle, mean MV cross-scatter intensity was calculated in a 100 × 100 pixel-wide region of interest (ROI) that was centered at the piercing point (i.e. projected location of focal spot in a projection). This process was repeated with default TrueBeam 1D antiscatter grid and with 2D grid to evaluate the effect of grid type on MV cross-scatter intensity. Ratio of cross-scatter intensities with default 1D grid and with 2D grid yielded the scatter rejection performance of 2D grid.
Likewise, scatter estimation error (SEE) of the GSS method was calculated for each projection pair as below,

$$\text{SEE} = 100 \times \frac{|\text{mean(} \text{GSS based MV scatter} - \text{reference scatter)}|}{\text{mean(} \text{reference MV cross scatter)}}. \quad (3)$$

2.5.2. Image domain evaluations
HU loss represents how HU values degrade in kV–MV images due to MV cross-scatter when compared to kV-only images. HU loss in kV–MV images was evaluated for default 1D grid, 2D grid and 2D grid + GSS configurations with their respective kV only counterparts. Specifically, we introduce $\Delta \text{HU}_{\text{method}}$ which represents the average absolute HU difference between any method’s output for kV–MV and their output for their kV only reconstruction. The formula is as follows:

$$\Delta \text{HU}_{\text{method}} = |\text{HU}_{\text{kV Only, method}} - \text{HU}_{\text{kV+MV, method}}|. \quad (4)$$

This approach assures that HU loss is solely due to MV cross-scatter for any given scatter suppression method evaluated.

HU loss is dependent on the ROI chosen for analysis. Thus, to reduce ROI selection bias in the analysis of HU loss, the full cross-section of the head-sized electron density phantom was used as the ROI for HU loss calculations. On the other hand, for anthropomorphic objects, the ROIs were chosen as circular neighborhoods with radii ranging from 7 pixels (for thorax phantom) to up to 15 (for pelvis phantom) as shown in figure 3.

Relative contrast-to-noise ratio improvement factor ($k\text{CNR}$) was calculated using the high contrast objects in both head-sized and pelvis-sized phantoms as ROIs and indicates the change in CNR in kV–MV images in comparison with CNR values with default 1D grid (Altunbas et al 2019).

The ROIs used for kCNR measurement, and their respective name and relative electron densities could be found in figure 4.

Finally, we introduced identity function profile (IFP) as a graphical representation of HU loss. Ideally, if all scatter due to MV cross-scatter were suppressed, pixel-by-pixel HU values would be the same for kV only and kV–MV reconstructions. Thus, if we were to plot a histogram between each corresponding pixel, the optimal scatter suppression method would result in an identity function. Hence, we could measure the optimality of any suggested scatter suppression method by evaluating how close to an identity function their profile resides.

Attenuation coefficient to HU conversion was calculated by measuring attenuation coefficients in water equivalent background section of the head-sized electron density phantom. This measurement was done using 2D grid only CBCT images acquired at 125 kVp, without the GSS correction method. Subsequently, the same water attenuation coefficient was used for HU conversion in 1D grid and 2D grid + GSS images. This process was repeated for CBCT scans acquired with BT filter. A separate HU conversion factor was not used for each grid type, because our goal was to show the relative change in attenuation coefficients in the 3 different scatter mitigation configurations we have investigated.
3. Results

3.1. Scatter suppression performance in projection domain

A comparison of average scatter values for different protocols is presented in figures 5(a)–(c). When compared to the default 1D grid in TrueBeam, 2D grid provided significant MV cross-scatter rejection, and reduced MV cross-scatter intensity by a factor of 3.19 ± 0.15 on the average across all protocols.

Figure 6 shows the SEE for each imaging protocol versus their respective MV cross-scatter to primary ratio (SPR) in the same ROI. As can be seen, SEE is relatively close to 0 in all scenarios with a mean of 6.97 ± 3.76% while cross-scatter estimation performance of the GSS method appears to be lower when MV SPR is below 0.5. This was in part attributed to inaccuracies in measuring reference (or ground truth) MV cross-scatter values. This issue was further elaborated in the Discussions section.

3.2. Scatter suppression performance in reconstructed image domain

Figure 7 shows the effect of MV cross-scatter on image quality in head-sized phantoms. The difference in images with and without MV beam demonstrates the HU degradation qualitatively. As expected, median HU loss with the default 1D grid was severe; HU loss was 182 for the case of 1200 cGy dose delivery with 6 MV beam and 10 × 10 cm² field size, while using the CBCT scan protocol with bowtie (BT) filter (figure 7(b)). MV field size and kV mAs were the two major factors that affected the HU loss (as observed by comparing the pair of figures 7(a) and (b)). When 2D grid was in use, HU loss, averaged across all protocols, was reduced from 38 HU to 16 HU.

When 2D grid was combined with GSS residual scatter correction, HU loss, averaged across all protocols, was further reduced to 4.8 HU (figure 7). Qualitatively, difference images (figure 7) also show that the effect of MV cross-scatter on HU accuracy was minimized when 2D grid + GSS approach was used. HU loss in 6 MV ad 10 MV beam deliveries were comparable, 3.2 HU versus 2.4 HU respectively, when 2D grid + GSS was used, indicating applicability of our method to different MV beam energies. When residual MV cross-scatter was present, ring artifacts were induced in images, particularly in images acquired with the 2D grid (figure 7). These artifacts were suppressed after cross-scatter correction with the GSS method.

A similar trend was observed in pelvis-sized phantom images (figure 8). Due to larger phantom size, MV SPR was higher, which yielded even larger HU degradation. The use of 2D grid reduced HU loss across all protocols from 92 HU to 58 HU. When 2D grid was combined with GSS, mean HU loss was further reduced to 9.3 HU. While reduction of MV beam dose from 1200 to 200 cGy led to proportionally lower HU loss, our method promptly recovered HU values in both cases; mean HU loss was reduced from 85 HU to 64 HU and from 64 HU to 10 HU in 2D grid only and 2D grid + GSS imaging protocols, respectively. Images acquired without bowtie filter (figures 8(c) and (d)) had lower kV imaging dose, and therefore, the intensity and detrimental effects of MV cross-scatter were relatively large in these image sets. While GSS method restored the HU values to a large extent, increased noise due to MV cross-scatter was visible in images. Increase in image noise was less pronounced in images acquired with bowtie filter and imaging dose (figures 8(a) and (b)).

HU loss as a function of imaging protocols is gathered in figure 9 which includes box plots of HU loss in (a) head and (b) pelvis sized phantoms. Each box plot consists of one box and two whiskers where the line within the box is median; upper and lower edges of the box correspond to 75th and 25th percentile and upper and lower
whiskers correspond to maximum and minimum data points. Across all imaging protocols investigated, 2D grid + GSS approach consistently led to substantial reduction in HU loss. Increasing the MV field size from $3 \times 3$ to $10 \times 10$ cm$^2$ caused the highest increase in MV cross-scatter fluence among head-sized phantoms, and hence, increased HU loss up to 182 HU with 1D grid. With 2D grid and 2D grid + GSS, median HU loss was reduced to 16 HU and 3 HU on the average, respectively. In images acquired without BT filter, HU loss was also

Figure 5. Measured MV cross-scatter intensity in projections as a function of imaging protocols and grid type. Cross-scatter intensities in head-sized and pelvis-sized electron density phantoms are shown in (a) and (b), respectively. Cross-scatter intensities measured with anthropomorphic thorax and pelvis phantoms are shown in (c). The central mark, box and whiskers correspond to median, 25th to 75th percentile and min/max, respectively. Using 2D grid instead of 1D grid reduces MV cross-scatter intensity by a factor of 3.2 on the average across all imaging protocols.
substantially larger, as well as visually apparent in CBCT images (figures 7(c), 8(c), (d)). Median and range of HU loss across all imaging protocols for all scatter suppression methods are summarized in table 1.

As in indicated in figures 10, 2D grid rejects a large amount of scatter which improved median CNR by a factor of 1.6 across all imaging protocols. Addition of GSS further improved CNR and resulted in a factor of 3.3 improvement in median CNR values across all protocols in comparison with the default 1D grid. Increase in CNR with GSS method was largely due to reduction of shading artifacts, and associated reduction in calculated noise in ROIs. Median and range of kCNR values across all imaging protocols are summarized in table 2.

Next, anatomically realistic pelvis and thorax phantom images are shown in figures 11(a) and (b), respectively. In pelvis phantom, median HU loss among ROIs reached 74 HU, when 2D grid was not used. The loss of HU accuracy was also evident in the difference images. With 2D grid, median HU loss decreased to 24.8 HU range. When 2D grid was combined with GSS residual scatter correction, HU loss was further reduced to 1.3 HU. Similarly for thorax, HU losses were up to 132 HU. When 2D grid was in use, HU loss decreased to 46 HU. When 2D grid was combined with GSS residual scatter correction, median HU loss was reduced to 3.5 HU. Difference images qualitatively show the high degree of agreement in HU values of kV-only and kV–MV acquisitions, when 2D grid + GSS approach was used.

Finally, IFPs of using 2D grid without and with GSS over pelvis and thorax are presented in figure 12. While 2D grid generates an IFP close to the ideal value (identity function) for both pelvis and thorax, there is a subset of pixels where a noticeable divergence from the ideal profile is evident, indicating that there are pixels whose HU values are not restored reasonably well (figures 12(a) and (c)). These pixels often correspond to high density regions, such as bony anatomy, where MV cross-scatter to primary ratio was higher in their projections.

On the other hand, 2D grid + GSS method provides much better agreement with the ideal profile (figures 12(b) and (d)), indicating superiority of 2D grid + GSS over 2D grid only approach. Quantitatively, in pelvis phantom, root mean square error (RMSE) with respect to the ideal grid property was equal to 54 and 25.2 HU for 2D grid only and 2D grid + GSS respectively. The same measure, when calculated for thorax changed from 49.2 HU in 2D grid only to 17.3 HU in 2D grid + GSS.

4. Discussion

In this work, we introduced a novel and robust method of rejecting and correcting MV cross-scatter in kV projections. A 2D antiscatter grid developed for kV imaging rejected MV cross-scatter with a mean energy of 300 keV (Taylor et al 1999) as indicated in figures 5, 9 and 10, thereby improving CT number accuracy and CNR. Specifically, CNR improvement is a significant advantage of cross-scatter rejection with 2D antiscatter grids, which cannot be achieved with scatter correction methods. However, MV cross-scatter fluence was not fully rejected by the 2D grid, and effects of residual MV cross-scatter incident on the FPD were still evident in CBCT images. As demonstrated in this work, our GSS method was efficient in correcting residual cross-scatter (figures 9 and 10), such that CT number accuracy in concurrent kV–MV beam delivery was comparable to...
kV-only CBCT images. HU accuracy and CNR improvement were comparable for 6 and 10 MV beams, indicating that beam energy plays a minor role in mitigating the effects of cross-scatter.

In addition to MV cross-scatter, MV x-rays due to linac head leakage were also part of the contaminant MV image signal in kV projections. Since our methods corrected all sources of contaminant x-rays regardless of their origin, effects of head leakage on kV projections were also expected to be reduced by our methods. However, performance evaluation of our methods in suppressing the effects of head leakage was not studied separately, an area of potential future research.

In contrast to kV-only scatter, the effects of MV cross-scatter strongly depend on the kV imaging dose, or kV primary signal. This is because, the fraction of MV cross-scatter in image signal is larger at lower kV doses, and vice versa. This effect was demonstrated in figures 8(b) and (c); images acquired with bowtie filter have twice the kV primary signal intensity (when compared to no bowtie filter scans), and MV cross-scatter to primary ratio was halved. Therefore, HU loss in images acquired with bowtie filter was less (figure 9(b)). While GSS method restores HU accuracy to a large extent, CNR loss may not be recovered, as evidenced by relatively noisy

**Figure 7.** kV CBCT images of head-sized electron density phantom acquired without and with concurrent MV beam delivery when (a) 10 MV beam with a $3 \times 3$ cm$^2$ field size and 1200 cGy dose in presence of bowtie filter (b) 6 MV beam with $10 \times 10$ cm$^2$ field size and 1200 cGy dose in presence of bowtie filter (c) 6 MV beam with a $3 \times 3$ cm$^2$ field size and 1200 cGy dose in absence of bowtie filter (d) 6 MV beam with a $3 \times 3$ cm$^2$ field size and 1200 cGy dose in presence of bowtie filter is applied. The third row shows the HU difference between the two CBCT scans acquired with and without concurrent MV beam delivery (i.e. the difference between row one and two). HU window ranges $[-250, 250]$ for CBCT images and $[-100, 300]$ for difference.
appearance in images acquired without bowtie filter (figure 8(c)). Thus, CNR loss and HU loss due to MV cross-scatter can be further reduced by using higher kV imaging dose, when feasible.

Besides HU and CNR loss, another drawback of residual cross-scatter is the induction of ring artifacts in reconstructed images. Such ring artifacts are particularly visible in figure 7(b), in images acquired with 2D grid. This is mostly due to reduced efficacy of flat-field correction in the presence of residual scatter, and hence, suboptimal correction of grid’s septal shadows (Altunbas et al 2021, Yu et al 2020). Since GSS suppresses residual scatter effectively, it also reduces the ring artifacts caused by cross-scatter. Similar artifacts were also observed when default 1D grid was in place, as seen in figures 7(a) and (b).

MV cross-scatter intensity estimated by the GSS method was in good agreement with the measured reference values, when MV SPR was above 0.5. However, the accuracy of GSS method appeared to deteriorate at lower MV SPR values. Our observations indicated that this issue stemmed from two different sources. First, reference MV cross-scatter intensity values were measured by subtracting kV-only projections from kV–MV projections, which was considered the ground truth. In kV–MV acquisitions, kV and MV beams were triggered asynchronously as in a clinical kV–MV beam delivery scenario. Such asynchronous triggering caused delivery of MV pulses during kV detector readout phase, which manifests itself as detector row-to-row variations in MV cross-scatter intensity, and stripe artifacts in projections. Such high spatial frequency signal variations might not be fully accounted for by the GSS method. This issue was apparent when MV beam pulses were triggered at lower frequencies, such as during 6 MV beam delivery with 200 MU (figure 8(d)). Second, it was assumed that the image signal difference between kV-only and kV–MV projections was equivalent to MV cross-scatter scatter intensity. This approach assumes that the kV image signal is identical for a given kV and kV–MV projection pair acquired at the same source angle. However, source angles in kV and kV–MV projection pairs in a CBCT acquisition can be different from each other by 0.2–0.3 degrees, causing different kV beam attenuation paths and kV image signal intensities in these two data sets. Moreover, kV tube output, and detector response cannot be

![Figure 8.](image-url)
kept identical in separate kV-only and kV–MV acquisitions. Such variations in kV image signal intensity ultimately affect the accuracy of MV cross-scatter measurement. Such inaccuracies were accentuated when MV SPR is low, or in other words, majority of kV–MV projection signal is composed of kV-only signals. Problems listed above are inherent limitations of measuring reference MV cross-scatter intensity, which, in return, affects the performance assessment of the GSS method in low MV SPR imaging conditions.

Most of the data in this work was acquired using 1200 MU and 3 × 3 cm² field size, to emulate beam delivery conditions in SBRT treatments. Based on our clinical experience, multi-leaf collimator (MLC) apertures vary largely during SBRT delivery, which would affect MV cross-scatter intensity. In addition, delivered MU per treatment beam often covers a wide range, depending on the treatment site, dose constraints, treatment planning techniques and dose prescription. Therefore, a more accurate assessment of MV cross-scatter suppression can be done in the future by using variety of clinical SBRT delivery scenarios and SBRT treatment plans.

Table 1. Median [min max] HU loss values across all concurrent kV–MV imaging protocols.

| Phantom type         | HU loss 1D grid | HU loss 2D grid | HU loss 2D grid + GSS |
|----------------------|-----------------|-----------------|-----------------------|
| Head-sized E. density| 38.2 [15.2 182.5] | 16.7 [5.6 86.0] | 3.6 [2.4 9.4]          |
| Pelvis-sized E. density | 92.8 [47.4 177.2] | 58.7 [32.6 140.5] | 9.3 [5.5 18.2]        |
| Pelvis                | 53.8 [23.4 74.0]  | 24.8 [1.1 70.4]  | 1.3 [0.3 5.9]          |
| Thorax                | 68.1 [46.4 132.2] | 46.3 [24.4 86.6] | 3.5 [1.5 7.2]          |
One of the concerns in using 2D antiscatter grids is the reduced primary signal, and its potential adverse effects on CNR. While antiscatter grids reduce primary fluence incident on the detector and increase noise, scatter rejection provided by the antiscatter grid improves contrast. Therefore, the overall change in CNR depends on the interplay between the CNR degradation caused by primary reduction and CNR improvement due to scatter rejection provided by the antiscatter grid. The 2D grid used in this study has a primary transmission fraction of 85%, whereas the 1D grid has primary transmission fraction of 70% (Altunbas et al 2019, 2017). As a result, CNR degradation due to primary beam attenuation by the antiscatter grid is less with the

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**Figure 10.** kCNR for (a) head-sized (b) pelvis-sized electron density phantoms as a function of imaging protocol. The central mark, box and whiskers correspond to median, 25th to 75th percentile, and min/max values, respectively.

**Table 2.** Median [min max] CNR improvement values (kCNR) with 2D grid and 2D grid + GSS across all concurrent kV–MV imaging protocols.

| Phantom type          | kCNR          | 2D grid | 2D grid + GSS |
|-----------------------|--------------|---------|--------------|
| Head-sized E. density | 1.8 [1.2 6.7] | 2.1 [1.2 13.0] |
| Pelvis-sized E. density | 1.4 [0.7 3.2] | 2.1 [0.8 3.8] |

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One of the concerns in using 2D antiscatter grids is the reduced primary signal, and its potential adverse effects on CNR. While antiscatter grids reduce primary fluence incident on the detector and increase noise, scatter rejection provided by the antiscatter grid improves contrast. Therefore, the overall change in CNR depends on the interplay between the CNR degradation caused by primary reduction and CNR improvement due to scatter rejection provided by the antiscatter grid. The 2D grid used in this study has a primary transmission fraction of 85%, whereas the 1D grid has primary transmission fraction of 70% (Altunbas et al 2019, 2017). As a result, CNR degradation due to primary beam attenuation by the antiscatter grid is less with the
Figure 11. kV CBCT images of (a) pelvis (b) thorax phantoms acquired without and with concurrent MV beam delivery. The third row shows the HU difference between the two CBCT scans acquired with and without concurrent MV beam delivery (i.e. the difference between row one and two). HU window range, pelvis phantom: \([-400\ 200]\) for CBCT images and \([-50\ 150]\) for difference. Thorax phantom: \([-450\ 150]\) for CBCT images and \([-50\ 100]\) for difference.

Figure 12. 2D HU correlation histograms between kV–MV and kV only CBCT images. Ideally, HU value for a given voxel would be identical in kV–MV and kV only images, and hence, only the histogram bins along the red line would be populated. HU histograms in (a) and (b) are generated from pelvis phantom images acquired with 2D grid only, and 2D grid + GSS, respectively. Likewise, HU histograms in (c) and (d) are generated from thorax phantom images. The color bar shows the histogram bin entries at given HU value.
2D grid. Moreover, the 2D grid provides a factor of 3.3–7.3 (depending on the phantom thickness) better scatter rejection than 1D grid, and associated CNR improvement due to scatter rejection is higher with 2D grid (Altunbas et al. 2017, 2019). Thus, due to higher primary transmission and efficient scatter rejection, CNR improvement by 2D grid can be achieved without increasing the imaging dose in head and pelvis sized phantoms as investigated in prior studies (Park et al. 2021). In this work, further CNR improvement with 2D grids was realized when kV and MV beams were used concurrently. This is due to substantially better MV cross-scatter rejection performance of 2D grids and reduction of image artifacts, when compared to 1D grids.

The 2D grid prototype used in this study has a substantially larger grid pitch than the grid pitch of the conventional 1D grid (2 mm versus 0.167 mm). There are two major reasons for using such a large grid pitch in the 2D grid. First, large grid pitch reduces the footprint of the tungsten septa on the detector, and thus improves primary transmission. Such large grid pitches are feasible to fabricate due to self-supporting structure of the 2D wall array. Second, our method for correcting any leftover scatter with the GSS method—as explained in section 2.2—requires clear definition of grid holes and grid wall shadows in projections. Thus, grid pitch is required to be larger than the pixel pitch.

Although MV cross-scatter rejection by 0.1 mm thick tungsten walls may seem counter intuitive, it can be justified by analyzing the energy spectrum of the MV cross-scatter (Taylor et al. 1999). The average energy of MV cross-scatter goes down as a function of exit angle (exit angle refers to the angle between the primary MV beam direction and the detector used to measure MV cross-scatter). While average energy of 6–10 MV primary beam is about 2–3 MeV, average energy of cross-scatter at 90 degree exit angle is about 240–300 keV. Another important detail is the quantum efficiency of the detector. While the scatter rejection performance of the 2D grid is reduced at higher energies, such high-energy scattered x-rays are less likely to interact in the detector and contribute to image signal. Such energy dependent response of the detector to MV cross-scatter can be potentially investigated in simulations in a future study. Moreover, MV cross-scatter rejection performance of our 2D grid was measured with respect to the 1D grid, which is composed of lead lamellae. Even though the linear attenuation coefficient of tungsten is reduced at 240 keV, similar behavior is also expected for the lead lamellae of the 1D grid. As a result, it is reasonable to expect better MV cross-scatter suppression performance from the 2D grid when compared to 1D grid.

Finally, we note how CBCT images acquired with 1D grid and bowtie filter show severe shading artifacts in the periphery of the phantom (figures 7 and 8). This is largely caused by the increased scatter-to-primary ratio in the periphery of the projections due to bowtie filter (Lazos and Williamson 2010, Altunbas et al. 2017). In clinical CBCT images, such shading artifacts are substantially less due to scatter correction. Whereas, in this work, additional scatter correction methods were not employed with the 1D grid. Scatter correction method in the clinical TrueBeam CBCT system was not used in conjunction with the default 1D grid, because all concurrent kV–MV imaging studies were done using Developer mode, where TrueBeam scatter correction option was not available.

5. Conclusion

Overall, a series of similar observations on different phantoms and imaging protocols indicate the strength of using the proposed 2D antiscatter grid and residual scatter correction method in mitigating the effects of MV cross-scatter in concurrent kV CBCT and MV beam delivery. HU loss due to MV cross-scatter was recovered to a large extent, and factor of 2–3 improvement in CNR was observed in concurrent kV CBCT scans and MV beam delivery. Unlike previously suggested methods, our methods do not require MV beam interruption or reduction in kV image acquisition rate, which makes them suitable for kV imaging during high dose-rate MV therapy delivery, such as SBRT, and for fast kV–MV image acquisition protocols, such as breath-hold CBCT.

While proposed methods were evaluated in the context of image guided external beam radiation therapy, our methods can potentially be used in other applications such as imaging during high dose-rate brachytherapy, and beyond radiation therapy, such as high x-ray energy industrial imaging and security imaging applications.

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