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Scatter Reduction and Correction for Dual Source Cone-beam CT using Pre-patient Grids

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Abstract

Purpose—Scatter significantly limits the application of the dual-source CBCT by inducing scatter artifacts and degrading CNR, Hounsfield-unit (HU) accuracy and image uniformity. Although our previously developed interleaved acquisition mode addressed the cross scatter between the two x-ray sources, it doubles the scanning time and doesn't address the forward scatter issue. This study aims to develop a pre-patient grid system to address both forward and cross scatter in the dual-source CBCT.

Methods—Grids attached to both x-ray sources provide physical scatter reduction during the image acquisition. Image data were measured in the unblocked region, while both forward and cross scatter were measured in the blocked region of the projection for post-scan scatter correction. Complementary projections were acquired with grids at complementary locations and were merged to form complete projections for reconstruction. Experiments were conducted with different phantom sizes, grid blocking ratios, image acquisition modes and reconstruction algorithms to investigate their effects on the scatter reduction and correction. The image quality improvement by the pre-patient grids was evaluated both qualitatively through the artifact reduction and quantitatively through CNR, HU accuracy and uniformity using a CATphan@504 phantom.

Results—Scatter artifacts were reduced by scatter reduction, and were removed by scatter correction method. CNR, HU accuracy and image uniformity were all substantially improved. The simultaneous acquisition mode achieved comparable CNR as the interleaved and sequential modes after scatter reduction and correction. Higher grid blocking ratio and smaller phantom size led to higher CNR for the simultaneous mode. The iterative reconstruction with TV regularization was

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Conflict of Interest

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more effective than the FDK method in reducing noise caused by the scatter correction to enhance CNR.

Conclusions—The pre-patient grid system is effective in removing the scatter effects in the simultaneous acquisition mode of the dual-source CBCT, which is useful for scanning time reduction or dual energy imaging.

Keywords

Dual energy imaging; Cross scatter; Grid blocking ratio; Scatter artifact; Total variation regularization; CNR enhancement; Simultaneous acquisition

I. Introduction

Cone-beam computed tomography (CBCT) using an x-ray source and a single flat panel detector has been widely used in a variety of applications in both radiology and radiation oncology.^{1–3} Recently, we developed and benchmarked a dual-source, dual-detector CBCT system, which has several advantages over the single source CBCT system.^{4, 5} The dual-source CBCT can provide dual-energy imaging without the use of fast kV switching.⁶ It may also reduce the total scan time required compared to current single source CBCT systems, and can be configured with two non-coplanar imaging chains to alleviate cone-beam artifacts.⁷ A dual-source onboard CBCT system would also allow for simultaneous, orthogonal radiographs or digital tomosynthesis, which have been shown to be highly useful for target localization in radiation therapy.^{8–10}

Scatter has been a major problem with kilovoltage (kV) CBCT due to its large imaging field size. Scattered radiation detected by the imager can severely degrade image quality by inducing significant scatter artifacts, degrading the CT number linearity, and reducing the contrast-to-noise ratio (CNR). The effects of scatter are even more severe for dual-source CBCT due to the cross scattering between the two x-ray sources. Many techniques have been proposed to address the scatter issue. These techniques can be generally divided into two major types: (1) direct scatter reduction during image acquisition;^{11–15} and (2) post scan scatter correction, in which scatter distributions are measured and/or modeled for each patient and subtracted out after the scan.^{16–21} However, the efficiency of direct scatter reduction methods is frequently limited, and often require increased dose to the patient. Post scan scatter correction methods do not address the scatter-related x-ray quantum noise and can potentially degrade the CNR.^{18, 22} In addition, the traditional scatter correction methods using beam-blocks typically require two scans to measure scatter and acquire image data separately. Consequently, patient dose is increased and the accuracy of scatter correction is limited because scatter was measured under a condition different from the image acquisition condition. Recently, Wang et al developed a moving blocker system^{23, 24}, and our group developed a synchronized moving grid system (SMOG)^{25, 26} to combine both scatter reduction and scatter correction in the system. Our preliminary study using SMOG on a single source CBCT system showed that it effectively removed the scatter artifacts and enhanced CNR while using similar imaging dose and scan time as a conventional CBCT system.²⁶ However, no such technique has been developed for the dual source CBCT system. We recently developed an interleaved acquisition mode for the dual source CBCT

system to avoid the cross scatter between the two x-ray sources.⁴ However, the method requires doubling of the scanning time, and does not address the forward scatter issue.

In this study, we will develop pre-patient grids for scatter reduction and correction for the dual source CBCT system to address both the cross scatter and forward scatter issue. Static grids were attached to both x-ray sources during the image acquisition of the dual source CBCT to provide physical scatter reduction. Similar to the SMOG system, image data were measured in the projection region unblocked by the grids, while both forward and cross scatter were measured in the projection region blocked by the grid for post scan scatter correction. Complementary projections for both sources were acquired through multiple scans with grids offset by a distance equal to the grid interspace after each scan, and were then merged to form complete projections for reconstruction. Experiments were conducted with different phantom sizes, grid blocking ratios, image acquisition modes and reconstruction algorithms to investigate their effects on the scatter reduction and correction. The image quality improvement by the pre-patient grids was evaluated both qualitatively through the artifact reduction and quantitatively through CNR, Hounsfield unit (HU) value accuracy and uniformity.

II. Materials and Methods

II.A. Experimental set up

II.A.1. Dual-source CBCT system—Our in-house developed dual-source CBCT system consists of two x-ray tubes (Varian G-242, Varian Medical System, Palo Alto, CA) and two flat panel detectors (PaxScan® 4030CB, Varian Medical System, Palo Alto, CA) positioned orthogonally to each other, as shown in Fig. 1.⁵ In this system, both tubes and detectors are fixed and the object being imaged is rotated by the rotational stage to acquire projections at different angles. The distance between each detector and the center of the rotational stage is 50cm, and the distance between each x-ray source and the center of the stage is 100cm, same as the set up in the clinical on-board imager (OBI) system on a Linac. The size of each flat-panel detector is 397×298 mm² with a pixel pitch of 194μm.

Bowtie filters for full fan scans were attached to both x-ray sources. Two pre-patient 1D grids made of 95WNIcu tungsten alloy were attached in front of the bow tie filters for scatter reduction and scatter measurement. The thickness of each grid is 3.5mm. Using the attenuation coefficient of tungsten at energy of 100 keV, it can be calculated that only 9.00E-14 of incident photons will pass through the 3.5 mm thick grid. Therefore, it's reasonable to assume that no primary photons will go through the blocked area of the grid. The grid blocking ratio is defined as the ratio of blocked area and unblocked area. In this study, grids with blocking ratios of both 1:1 and 2:1 were used to investigate the effects of grid blocking ratio on the scatter reduction and scatter correction. Same as in our previous studies⁴, 120 kVp and 1mAs (40mA and 25ms) per projection were used, and 554 projections per detector were acquired evenly spaced over 360°, giving a total of 1108 projections over 360°.

II.A.2. Acquisition modes—Three acquisition modes were investigated in our study: simultaneous mode, interleaved mode and sequential mode. In the simultaneous mode, the

two x-ray tubes fire simultaneously and the two detectors record at the same time at each scan angle during a one-rotation scan. In the interleaved acquisition mode, the tubes fire alternatively at each scan angle with the rotational speed of the stage reduced to half. In the sequential mode, the two x-ray tubes fire sequentially in two separate scans. The simultaneous mode has the shortest scanning time as well as the most significant scatter effects among the three acquisition modes. Details of the three acquisition modes were described in our previous studies.⁴

II.B. Scatter reduction, correction, penumbra derivation and projection merging

The grids attached to the x-ray sources provide physical reduction of the scatter by blocking part of the beams. The post scan scatter correction, penumbra derivation and projection merging process is implemented in five steps as shown in Fig. 2: (1) acquire blank scan (without phantom) and phantom scan with grids; (2) estimate the scatter distribution based on the measured scatter data from blocked areas in each blank scan and phantom scan projection using a cubic spline interpolation/extrapolation along the longitudinal direction; (3) subtract the scatter distribution from projection images for scatter correction; (4) negative logarithm transformation of the division of the scatter-corrected phantom scan projections and the corresponding scatter-corrected blank scan projections to derive image information in both the unblocked region and the penumbra region of the grids; (5) complementary projections are acquired at each angle with the grid offset by a distance equal to the grid interspace after each exposure, and complete projection images are obtained by merging the scatter-corrected complementary projections. Note that step 4 derives the image information in the penumbra region of the grid by dividing the phantom scan by the blank scan in the penumbra region. To circumvent the “division by zero” problem in regions where the blank scan signal is minimal under the grid penumbra, we defined a “dead area” as a region with intensity less than a cutoff value, e.g. less than 10% of the blank scan signal in the unblocked region. The dead areas were masked during the division process. Deriving image information in the penumbra region effectively reduced the blocked regions in the projections, and is a key step for achieving complete projections without blocked regions after projection merging, as demonstrated in our previous studies.^{25, 26}

Due to the unavailability of a motor to automatically move the grid back and forth at each angle, the grid was static during each full rotation scan, and was manually moved to complementary positions in-between scans in our experiments. The number of the scans needed is determined by the grid blocking ratio. For example, the 1:1 grid blocks half of the x-ray beam in each projection and two projections are needed at each angle to obtain complete information. As a result, two scans are needed with the grid off set in-between scans.

The effect of scatter reduction only was also studied using the partially blocked projections directly for reconstruction without the scatter correction process. Basically, the scatter reduction only method uses the same process as shown in Fig. 2 except skipping step 2 and step 3.

II.C. Reconstruction algorithms

Both the Feldkamp, Davis, and Kress (FDK) back projection method and the iterative reconstruction with total variation (TV) method were used for CBCT image reconstruction to evaluate their effects on the image quality after scatter reduction and correction.^{27, 28} The FDK method reconstructs images by directly back project the filtered projections. The TV-based iterative reconstruction used in this study takes an adaptive approach proposed in the ASD-POCS algorithm developed by Sidky and Pan.²⁸ A relative weighting factor between TV minimization and the data fidelity optimization is chosen as the initial starting point, and the algorithm will automatically and adaptively adjust the weighting factor during iterations to balance the effects of data fidelity optimization and TV minimization. The effectiveness of the algorithm is partially determined by the initially chosen relative weighting factor. However, the sensitivity of the final results to this weighting factor is less than that in the traditional TV iterative reconstruction methods with a fixed relative weighting factor for the TV minimization. In general, the FDK method has high computation efficiency, but is prone to noise and artifacts caused by under-sampling. The iterative reconstruction with TV regularization has low computation efficiency, but is effective in reducing the noise or streak artifacts caused by under-sampling in the reconstructed images. However, the TV regularization term may also lead to blurriness of the image.

II.D. Evaluation methods

The CATphan ® 504 (PhantomLab, Salem, NY) commonly used in CT QA was used in this study to evaluate the image quality improvement with scatter reduction and correction. The size of the phantom was varied by attaching different annuli outside the CTP404 module of the CATphan to study the effects of the phantom size on the scatter reduction and correction efficiency. In this study, phantoms with the diameters of 15cm, 20cm and 30cm were investigated.

The effectiveness of scatter reduction and scatter correction was analyzed qualitatively by evaluating the artifact reduction and quantitatively by evaluating the CNR, the accuracy of reconstructed HU values and the uniformity of reconstructed images. The CNR defined in our previous study was measured for four inserts of different contrasts in the CTP404 module using regions of interests (ROIs) drawn both in the inserts (red circles) and in the region adjacent to the inserts (yellow circles), as shown in Fig. 3.²⁶

To evaluate the HU accuracy, the reconstructed pixel values were converted to HU values using a linear regression between mean reconstructed values of the eight inserts in Fig. 3 and their corresponding nominal HU values. The uniformity was evaluated in the uniformity module of the CatPhan. The uniformity index was obtained by calculating the standard deviation of the mean values of five ROIs (20*20 pixels) located in the center and boundary of the phantom, as shown in Fig. 4.

III. Results

III.A. Scatter correction and projection merging

The profiles of the center column in a projection image are shown in Fig. 5, to illustrate the scatter measurement, estimation and correction process for blank scan and phantom scan projections. To avoid the penumbra effect of the grid, only the measured data inside the central one third of each blocked region are used for scatter estimation. The average of these scatter data is used as the sampled scatter value at each blocked region, as shown in Fig. 5. A cubic spline interpolation/extrapolation is carried out to estimate the entire scatter profile based on the sampled scatter values at different blocked regions in the projection.

The negative logarithm transformation was then applied to the division between the scatter-corrected phantom scan projections and the corresponding scatter-corrected blank scan projections to derive the image information in the partially blocked projection. Complementary partially blocked projections were added together afterwards to obtain a complete projection at each scan angle. This process was repeated for different scan angles to obtain a complete set of scatter corrected projections for CBCT reconstruction. Fig. 6 shows an example of projection merging between two complementary projections.

III.B. Effects of acquisition modes on scatter reduction and correction

Fig. 7 shows the images reconstructed by the FDK method using scatter reduction and correction for different acquisition modes. As indicated by the arrows, the scatter artifacts in the original images have been substantially reduced after scatter reduction and correction. Table 1 shows the change of CNR after scatter reduction and scatter correction for different acquisition modes. Four ROIs of different contrasts in Fig. 3 were used to evaluate the effects.

III.C. Effects of phantom size on scatter reduction and correction

Fig. 8 shows the effects of phantom size on the scatter reduction and correction for the simultaneous acquisition mode. The FDK method was used for reconstruction. Scatter artifacts, such as streak artifacts indicated by the arrows, were reduced after scatter reduction and removed after scatter correction. Note that the bright ring in the peripheral regions of the images for the 30cm-diameter phantom in Fig. 8 is caused by truncation artifacts.^{29, 30}

Table 2 shows the change of CNR after scatter reduction and scatter correction for different phantom sizes in the simultaneous acquisition mode.

III.D. Effects of grid blocking ratio on scatter reduction and correction

Fig. 9 shows the effects of grid blocking ratio on the scatter reduction and correction for the simultaneous acquisition mode using the FDK reconstruction method. As indicated by the arrows, the scatter induced artifacts were substantially reduced by scatter reduction only and were removed after correction using both grids. Table 3 shows the change of CNR after scatter reduction and correction using 1:1 and 2:1 grids in the simultaneous acquisition mode.

III.E. Effects of reconstruction algorithm on scatter reduction and correction

The results above were all generated by FDK reconstruction algorithm. Noise was increased in the scatter correction process, degrading the CNR especially for large phantom. Fig. 10 shows the comparison of CBCT images reconstructed by different algorithms after scatter reduction and correction in the simultaneous acquisition mode. As shown in the figure, noise was reduced substantially by the iterative reconstruction method. Table 4 shows the CNR of the inserts reconstructed by different algorithms.

III.F. Improvement of the HU accuracy by scatter reduction and correction

Table 5 shows the calibrated HU values in the reconstructed CBCT images of a 20cm-diameter phantom acquired with the 1:1 grid and simultaneous acquisition mode.

III.G. Improvement of the image uniformity by scatter reduction and correction

Table 6 shows the uniformity index of the reconstructed CBCT images after scatter reduction and scatter correction. A phantom of 20cm diameter, 1:1 grid and simultaneous acquisition mode were used for this study.

IV. Discussion

IV.A. Effects of scatter reduction and correction for dual-source CBCT

All the above studies showed that scatter artifacts in the CBCT acquired in the simultaneous acquisition mode were reduced by scatter reduction, and were removed by the scatter correction method. Scatter reduction alone was shown to enhance CNR, while adding scatter correction may enhance or degrade CNR due to the increase of both the contrast and the noise in the scatter subtraction process. Note that in Table 1, high density inserts, such as ROI3 (HU=990) and ROI4 (HU=340), had more decrease of CNR after scatter correction alone in the sequential mode than low density inserts, such as ROI1 (HU= -1000) and ROI2 (HU= -35). This may be due to the fact that high density inserts generate more scatter and correspondingly higher noise in the scatter signal. As a result, the effect of noise enhancement is more prominent during the scatter subtraction process leading to poorer CNR for high density inserts in the reconstructed images.

The HU accuracy was improved by both scatter reduction and scatter correction. Combining scatter reduction and scatter correction achieved the best accuracy for HU values. The image uniformity was also improved by both scatter reduction and correction due to the reduction and removal of the cupping artifacts caused by scatter, respectively.

IV.B. Effects of the acquisition modes of the dual-source CBCT

Comparing different acquisition modes, the original images from the simultaneous acquisition mode had the lowest CNR due to the cross scatter between the two x-ray sources. Interleaved mode had similar CNR as the sequential mode due to the avoidance of cross scatter during the scan. For all three modes, CNR increased after the scatter reduction, but slightly decreased for some inserts after the scatter correction alone due to the increase of noise in the scatter subtraction process. The simultaneous mode had the largest enhancement of CNR, and achieved similar CNR as interleaved and sequential modes after scatter

reduction and correction. This showed that the pre-patient grid system was effective in reducing and correcting both forward scatter and cross scatter in the simultaneous acquisition mode, which is the most preferable among the three acquisition modes due to its scanning efficiency.

IV.C. Effects of the phantom size

Regarding the effects of the phantom size, CNR decreased substantially as phantom size increased due to the increase of scatter, as shown in Table 2. Similar as in Table 1, scatter reduction increased the CNR while scatter correction decreased the CNR slightly for some inserts. Combining scatter reduction and correction led to enhancement of the CNR. Note that even though scatter reduction and correction can effectively remove the scatter to obtain similar contrasts for different phantom sizes, the CNR after scatter reduction and correction for a larger phantom was still substantially lower than that for a smaller phantom. This is because the higher scatter signal generated by the larger phantom corresponds to higher magnitude of noise, which leads to higher noise after scatter subtraction and therefore lower CNR in the scatter corrected CBCT images.

IV.D. Effects of the grid blocking ratio

Regarding the effects of the grid blocking ratio, results showed that higher grid blocking ratio achieved substantially more enhancement of CNR than lower grid blocking ratio due to the more drastic physical reduction of the scatter. It can also be intuitively understood that grids with a high grid blocking ratio blocks out more primary beams and shapes the cone beam into multiple fan beams, which effectively reduces the scatter and enhances CNR. However, on the other hand, higher grid blocking ratio requires more complex imaging procedures and image processing to obtain complete projection information of the patient. Thus, higher grid blocking ratio may be more preferable for imaging larger patients with severe scatter effects to balance between efficiency and image quality in the dual-source CBCT.

The effects of phantom size and grid blocking ratio can also be explained mathematically as following: Denote variables S and P as the mean values of the detected scatter and primary signals after scatter reduction by the grid, respectively. Denote variable n_c as the noise of the line integral image after scatter reduction and correction. Then the variance of n_c can be calculated as follows, as explained in the paper published by Zhu et al²²:

$$var(n_c) \approx \frac{1}{P} \left(1 + \frac{S}{P}\right)$$

A smaller phantom or higher grid blocking ratio will lead to lower S/P due to the smaller imaging scatter volume. Based on this equation, it will lead to lower $var(n_c)$ and higher CNR in the reconstructed images after scatter reduction and correction, which is consistent with the results and discussion above.

IV.E. Effects of reconstruction algorithms

Results showed that the iterative reconstruction with TV regularization achieved substantially higher CNR than the FDK method in the original CBCT images acquired without grids due to the noise-suppression by the TV algorithm. The enhancement of CNR by scatter reduction and correction was also more dramatic for the iterative method than the FDK method.

IV.F. Limitation of the current study

In this study, both grids were attached statically to the two x-ray sources during the scan due to unavailability of motors to move the grid during the scan. As a result, multiple scans were acquired to obtain complete projection information, which limited the scanning efficiency of the dual-source CBCT system. In the future, dual motors will be developed to move both grids back and forth at each angle to acquire complete projections during one scan, similar as that proposed in the SMOG system.²⁶ With the moving grid system, the dual-source CBCT can potentially reduce the scanning time to half of that of a single-source CBCT or perform dual energy imaging within one scan with minimal scatter effects from both forward and cross scatter.

V. Conclusion

A method using pre-patient grids has been developed for scatter reduction and scatter correction for a dual-source CBCT system. Results showed that the method effectively removed scatter artifacts, enhanced CNR, and improved the HU accuracy and image uniformity in the simultaneous acquisition mode. The simultaneous mode achieved comparable CNR as the interleaved and sequential modes after the scatter reduction and correction. Higher grid blocking ratio leads to more physical scatter reduction and as a result less scatter artifacts and higher CNR. Larger phantom size leads to more scatter generated and lower CNR after scatter reduction and correction. The iterative reconstruction with TV regularization was also effective in reducing the noise caused by the scatter correction algorithm to enhance the CNR.

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Abbreviations

CBCT	Cone-beam CT
kV	Kilovoltage
CNR	Contrast-to-noise ratio
SMOG	Synchronized moving grid system
HU	Hounsfield unit

OBI	On-board imager
FDK	Feldkamp, Davis, and Kress
TV	Total variation
ROI	Regions of interests

References

1. Jaffray DA, Siewerdsen JH. Cone-beam computed tomography with a flat-panel imager: initial performance characterization. *Medical physics*. 2000; 27:1311–1323. [PubMed: 10902561]
2. Siewerdsen JH, Moseley DJ, Burch S, Bisland SK, Bogaards A, Wilson BC, Jaffray DA. Volume CT with a flat-panel detector on a mobile, isocentric C-arm: pre-clinical investigation in guidance of minimally invasive surgery. *Medical physics*. 2005; 32:241–254. [PubMed: 15719975]
3. Xing L, Siebers J, Keall P. Computational challenges for image-guided radiation therapy: framework and current research. *Seminars in radiation oncology*. 2007; 17:245–257. [PubMed: 17903702]
4. Giles W, Bowsher J, Li H, Yin FF. Interleaved acquisition for cross scatter avoidance in dual cone-beam CT. *Medical physics*. 2012; 39:7719–7728. [PubMed: 23231319]
5. Li H, Giles W, Bowsher J, Yin FF. A dual cone-beam CT system for image guided radiotherapy: initial performance characterization. *Medical physics*. 2013; 40:021912. [PubMed: 23387760]
6. Johnson TR, Krauss B, Sedlmair M, Grasruck M, Bruder H, Morhard D, Fink C, Weckbach S, Lenhard M, Schmidt B, Flohr T, Reiser MF, Becker CR. Material differentiation by dual energy CT: initial experience. *European radiology*. 2007; 17:1510–1517. [PubMed: 17151859]
7. Forthmann P, Grass M, Proksa R. Adaptive two-pass cone-beam artifact correction using a FOV-preserving two-source geometry: a simulation study. *Medical physics*. 2009; 36:4440–4450. [PubMed: 19928075]
8. Jin JY, Yin FF, Tenn SE, Medin PM, Solberg TD. Use of the BrainLAB ExacTrac X-Ray 6D system in image-guided radiotherapy. *Medical dosimetry: official journal of the American Association of Medical Dosimetrists*. 2008; 33:124–134. [PubMed: 18456164]
9. Zhang Y, Yin FF, Segars WP, Ren L. A technique for estimating 4D-CBCT using prior knowledge and limited-angle projections. *Medical physics*. 2013; 40:121701. [PubMed: 24320487]
10. Ren L, Zhang Y, Yin FF. Medical Physics Letter: A limited-angle intrafraction verification (LIVE) system for radiation therapy. *Medical physics*. 2014; 41:020701. [PubMed: 24506590]
11. Siewerdsen JH, Jaffray DA. Optimization of x-ray imaging geometry (with specific application to flat-panel cone-beam computed tomography). *Medical physics*. 2000; 27:1903–1914. [PubMed: 10984236]
12. Siewerdsen JH, Moseley DJ, Bakhtiar B, Richard S, Jaffray DA. The influence of antiscatter grids on soft-tissue detectability in cone-beam computed tomography with flat-panel detectors. *Medical physics*. 2004; 31:3506–3520. [PubMed: 15651634]
13. Graham SA, Moseley DJ, Siewerdsen JH, Jaffray DA. Compensators for dose and scatter management in cone-beam computed tomography. *Medical physics*. 2007; 34:2691–2703. [PubMed: 17821977]
14. Sorenson JA, Niklason LT, Knutti DF. Performance characteristics of improved antiscatter grids. *Medical physics*. 1980; 7:525–528. [PubMed: 7421763]
15. Stankovic U, van Herk M, Ploeger LS, Sonke JJ. Improved image quality of cone beam CT scans for radiotherapy image guidance using fiber-interspaced antiscatter grid. *Medical physics*. 2014; 41:061910. [PubMed: 24877821]
16. Zhu L, Xie Y, Wang J, Xing L. Scatter correction for cone-beam CT in radiation therapy. *Medical physics*. 2009; 36:2258–2268. [PubMed: 19610315]
17. Ning R, Tang X, Conover D. X-ray scatter correction algorithm for cone beam CT imaging. *Medical physics*. 2004; 31:1195–1202. [PubMed: 15191309]

18. Siewerdsen JH, Daly MJ, Bakhtiar B, Moseley DJ, Richard S, Keller H, Jaffray DA. A simple, direct method for x-ray scatter estimation and correction in digital radiography and cone-beam CT. *Medical physics*. 2006; 33:187–197. [PubMed: 16485425]
19. Wagner FC, Macovski A, Nishimura DG. Dual-energy x-ray projection imaging: two sampling schemes for the correction of scattered radiation. *Medical physics*. 1988; 15:732–748. [PubMed: 3185410]
20. Zhu L, Bennett NR, Fahrig R. Scatter correction method for X-ray CT using primary modulation: theory and preliminary results. *IEEE transactions on medical imaging*. 2006; 25:1573–1587. [PubMed: 17167993]
21. Thing RS, Mainegra-Hing E. Optimizing cone beam CT scatter estimation in egs_cbct for a clinical and virtual chest phantom. *Medical physics*. 2014; 41:071902. [PubMed: 24989380]
22. Zhu L, Wang J, Xing L. Noise suppression in scatter correction for cone-beam CT. *Medical physics*. 2009; 36:741–752. [PubMed: 19378735]
23. Ouyang L, Song K, Wang J. A moving blocker system for cone-beam computed tomography scatter correction. *Medical physics*. 2013; 40:071903. [PubMed: 23822440]
24. Wang J, Mao W, Solberg T. Scatter correction for cone-beam computed tomography using moving blocker strips: a preliminary study. *Medical physics*. 2010; 37:5792–5800. [PubMed: 21158291]
25. Jin JY, Ren L, Liu Q, Kim J, Wen N, Guan H, Movsas B, Chetty IJ. Combining scatter reduction and correction to improve image quality in cone-beam computed tomography (CBCT). *Medical physics*. 2010; 37:5634–5644. [PubMed: 21158275]
26. Ren L, Yin FF, Chetty IJ, Jaffray DA, Jin JY. Feasibility study of a synchronized-moving-grid (SMOG) system to improve image quality in cone-beam computed tomography (CBCT). *Medical physics*. 2012; 39:5099–5110. [PubMed: 22894435]
27. Feldkamp L, Davis L, Kress J. Practical cone-beam algorithm. *Journal of the Optical Society of America A*. 1984; 1:612–619.
28. Sidky EY, Pan X. Image reconstruction in circular cone-beam computed tomography by constrained, total-variation minimization. *Physics in medicine and biology*. 2008; 53:4777–4807. [PubMed: 18701771]
29. Mawlawi O, Erasmus JJ, Pan T, Cody DD, Campbell R, Lonn AH, Kohlmyer S, Macapinlac HA, Podoloff DA. Truncation artifact on PET/CT: impact on measurements of activity concentration and assessment of a correction algorithm. *AJR American journal of roentgenology*. 2006; 186:1458–1467. [PubMed: 16632745]
30. Yan H, Wang X, Shi F, Bai T, Folkerts M, Cervino L, Jiang SB, Jia X. Towards the clinical implementation of iterative low-dose cone-beam CT reconstruction in image-guided radiation therapy: cone/ring artifact correction and multiple GPU implementation. *Medical physics*. 2014; 41:111912. [PubMed: 25370645]

Figure 1.
Experimental set up of the dual source CBCT system

Figure 2.

An example to illustrate the scatter correction and projection merging procedure using grid of blocking ratio 1:1.

Figure 3. Axial view of the CTP404 module with inserts of different contrasts. The red circles are the ROIs inside the inserts while the yellow ones are the ROIs surrounding the inserts.

Figure 4. Axial view of the conventional CBCT of the CTP486 uniformity module of the CATphan. Five ROIs were defined (one is in the center and the other four are at the edge) to calculate the uniformity of the image.

Figure 5.

An example illustrating the scatter estimation and correction process in (a). a blank scan and (b). a phantom scan. The blue solid line is the original 1 D profile in the longitudinal direction in a blank/phantom scan projection. The star dots are the measured scatter in the blocked areas. The black line is the interpolated scatter distribution. The red solid line is final profile after scatter correction by subtracting the black line from the blue line.

Figure 6. Profiles extracted from complementary projections acquired using 1:1 grid after blank scan normalization and negative logarithm transformation. (a) The red and blue solid lines were from two complementary projections acquired with the grid shifted by one grid inter-space. (b) Profile of the merged projection.

Figure 7. Reconstructed images of a CATphan of 20cm diameter using 1) simultaneous, 2) interleaved and 3) sequential acquisition mode. 2:1 grid was applied for scatter reduction and scatter correction. From left to right column: (a) without grid, (b) scatter reduction only and (c) scatter reduction and correction. Window width of 1000HU and window level of -200HU were used for all images. The FDK method was used for reconstruction.

Figure 8.

Reconstructed images of a CATphan of 15, 20 and 30cm diameters (from 1st to 3rd row) acquired by simultaneous mode (dual-source). 1:1 grid was applied for scatter reduction and scatter correction. From left to right column: (a) without grid, (b) scatter reduction only and (c) scatter reduction and correction. Display for 1st and 2nd : window width of 1000HU and window level of -200HU were used. Display for 3rd : window width of 2200HU and window level of 100HU were used. The FDK method was used for reconstruction.

Figure 9. Reconstructed images of a CATphan of 20 cm diameter acquired by simultaneous mode (dual-source). 1:1(1st row) and 2:1grid (2nd row) was applied for scatter reduction and scatter correction. From left to right column: (a) without grid, (b) scatter reduction only and (c) scatter reduction and correction. Window width of 1000HU and window level of -200HU were used for all images. The FDK method was used for reconstruction.

Figure 10.

Reconstructed images of a CATphan of 20cm diameter acquired by simultaneous mode (dual-source). 1:1 grid was applied for scatter reduction and scatter correction. Images were reconstructed from the FDK method (1st row) and iterative method (2nd row). From left to right column: (a) without grid, (b) scatter reduction only and (c) scatter reduction and correction. Window width of 1000HU and window level of -200HU were used for all images.

Table 1

Comparison between CNR acquired by simultaneous, interleaved and sequential modes. Phantom of 20cm diameter and 2:1 grid were used in the scan.

CNR (20 cm phantom, 2:1 grid)				
ROI	Acquisition Mode	Original Data	Scatter Reduction Only	Scatter Reduction and Correction
1	Simultaneous	58.3	95.0	95.4
	Interleaved	89.7	97.6	101.7
	Sequential	77.2	97.0	97.4
2	Simultaneous	9.9	12.6	12.4
	Interleaved	10.7	12.3	12.4
	Sequential	11.6	12.2	12.0
3	Simultaneous	37.8	64.8	66.2
	Interleaved	60.3	66.4	68.5
	Sequential	58.3	81.9	73.8
4	Simultaneous	14.9	21.9	20.0
	Interleaved	19.1	21.5	19.4
	Sequential	21.4	21.6	20.4

Table 2

Comparison between CNR in CBCT images acquired for various phantom sizes

CNR (15, 20 and 30cm phantom, simultaneous mode, 1:1 grid)				
ROI	Phantom Size: Diameter (cm)	Original Data	Scatter Reduction Only	Scatter Reduction and Correction
1	15	73.9	90.1	88.3
	20	53.4	63.5	66.9
	30	12.1	16.4	14.5
2	15	20.3	21.5	20.6
	20	8.8	10.1	9.1
	30	1.1	2.0	1.7
3	15	95.6	102.1	99.8
	20	48.5	58.2	57.5
	30	8.2	12.2	10.4
4	15	32.2	38.7	35.1
	20	14.4	18.5	18.1
	30	2.9	3.4	3.2

Table 3

Effects of scatter reduction and correction for different grid blocking ratios

CNR (20 cm phantom, simultaneous mode, 1:1 and 2:1 grid)				
ROI	Grid Blocking Ratio	Original Data	Scatter Reduction Only	Scatter Reduction and Correction
1	1:1	55.8	63.5	66.9
	2:1		95.0	117.6
2	1:1	9.3	10.1	9.1
	2:1		12.6	11.7
3	1:1	43.2	50.0	57.5
	2:1		64.8	66.6
4	1:1	14.6	18.5	18.1
	2:1		21.9	21.3

Table 4

Effects of reconstruction algorithms on scatter reduction and correction

CNR (20 cm phantom, simultaneous mode, 1:1 grid)				
ROI	Reconstruction Method	Original Data	Scatter Reduction Only	Scatter Reduction and Correction
1	FDK	53.4	63.5	66.9
	TV	74.4	84.5	117.0
2	FDK	8.8	10.1	9.1
	TV	10.6	19.1	19.4
3	FDK	48.5	50.0	57.5
	TV	57.5	67.0	91.2
4	FDK	14.4	18.5	18.1
	TV	23.5	25.6	25.5

Table 5

HU accuracy in reconstructed CBCT images of the 20cm-diameter phantom acquired with the 1:1 grid and the simultaneous scanning mode

ROI	Nominal HU	Reconstructed HU		
		Original	Scatter Reduction	Scatter Reduction and Correction
Air	−1000	−1041	−1008	−999
Polystyrene	−35	−23	−39	−33
LDPE	−100	−44	−120	−113
PMP	−200	−148	−191	−198
Teflon	990	960	971	991
Delin	340	320	359	347
Acrylic	120	97	135	122

Table 6

Uniformity index of three image sets: original CBCT, CBCT reconstructed with scatter reduction only and CBCT reconstructed with scatter reduction and correction. The phantom of 20 cm diameter, 1:1 grid and simultaneous mode were used.

Uniformity Index		
Original Data	Scatter Reduction Only	Scatter Reduction and Correction
27.71	17.18	9.49