The influence of bowtie filtration on cone-beam CT image quality

N. Mail
Radiation Medicine Program, Princess Margaret Hospital, Toronto, Ontario M5G 2M9, Canada and Ontario Cancer Institute, University Health Network, Toronto, Ontario M5G 2M9, Canada

D. J. Moseley
Radiation Medicine Program, Princess Margaret Hospital, Toronto, Ontario M5G 2M9, Canada; Ontario Cancer Institute, University Health Network, Toronto, Ontario M5G 2M9, Canada; and Department of Radiation Oncology, University of Toronto, Toronto, Ontario M5G 2M9, Canada

J. H. Siewerdsen and D. A. Jaffray
Radiation Medicine Program, Princess Margaret Hospital, Toronto, Ontario M5G 2M9, Canada; Ontario Cancer Institute, University Health Network, Toronto, Ontario M5G 2M9, Canada; and Department of Radiation Oncology and Department of Medical Biophysics, University of Toronto, Toronto, Ontario M5G 2M9, Canada

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The large variation of x-ray fluence at the detector in cone-beam CT (CBCT) poses a significant challenge to detectors’ limited dynamic range, resulting in the loss of skinline as well as reduction of CT number accuracy, contrast-to-noise ratio, and image uniformity. The authors investigate the performance of a bowtie filter implemented in a system for image-guided radiation therapy (Elekta oncology system, XVI) as a compensator for improved image quality through fluence modulation, reduction in x-ray scatter, and reduction in patient dose. Dose measurements with and without the bowtie filter were performed on a CTDI Dose phantom and an empirical fit was made to calculate dose for any radial distance from the central axis of the phantom. Regardless of patient size, shape, anatomical site, and field of view, the bowtie filter results in an overall improvement in CT number accuracy, image uniformity, low-contrast detectability, and imaging dose. The implemented bowtie filter offers a significant improvement in imaging performance and is compatible with the current clinical system for image-guided radiation therapy. © 2009 American Association of Physicists in Medicine. [DOI: 10.1118/1.3017470]

Key words: bowtie filter, Elekta synergy CBCT system, image quality, dose

I. INTRODUCTION

Cone-beam computed tomography (CBCT) systems are employed in radiation therapy to provide information regarding daily target and normal tissue localization during the course of fractionated radiation treatment. CBCT allows radiation to be directed at tumors with greater accuracy and precision than was previously possible. However, improving CBCT image quality further will allow more precise dose delivery to the tumor. Precision in tumor localization and reduction in setup error is based on image quality of CBCT system. However, flat-panel detectors used in CBCT have limited dynamic range compared to CT. These limitations are exposed by the large variation of x-ray fluence at the detector across the imaged field-of-view. It is often the case that the techniques used overwhelm the signal range of the detector at the periphery of the patient, leading to a loss of information in projections and artifacts in reconstruction due to the truncation of anatomy. As a result, effects include the loss of skinline and reduction in CT number accuracy, contrast, and image uniformity. The purpose of this study is to investigate the relative advantages and limitations of a bowtie filter for improving image quality of a commercially used CBCT Elekta Synergy System. The implementation of a bowtie filter in CBCT offers the method of addressing the issue of x-ray flux variation across the detector.

Previously, beam shaping x-ray filters or compensators have been used in computed tomography (CT) to modify the distribution of x-ray flux across the field of view. Compensator filters have been used in CT for beam hardening and reduction in dose to the patient. These compensators have been used in CBCT for scatter and heel compensation. Initially, compensators have been used to improve the detectability in film radiographs with the delivery of more uniform fluence at the film.

Bowtie filters potentially play a larger role in scatter, beam hardening, uniform fluence across the detector, and dose reduction. Quantitative investigations are performed on the Elekta Synergy radiotherapy image-guidance system. These investigations provides quantitative evidence of bowtie filter for improving skinline, image uniformity, CT number accuracy, contrast, and reduction in streak artifacts and patient dose.

II. MATERIALS AND METHODS

II.A. Cone-beam CT system

Investigations of bowtie filter use in CBCT were performed on Elekta oncology system shown in Fig. 1(a). The x-ray source uses a rotating anode x-ray tube (DunleeD604, IL) with maximum potential of 150 kVp, 14 deg tungsten
target, and 0.8 mm focal spot. The detector used was the RID1640-Al1 (Perkin Elmer, Wiesbaden, Germany) indirect-detection flat-panel imager with a 1024×1024 array of 0.4 mm×0.4 mm pixels and a 0.55 mm thick CsI:Tl x-ray converter. The geometry of the system was configured to have a source-to-isocenter distance of 100 cm and source-to-detector distance of 153.6 cm.

The x-ray tube and flat-panel imager were placed orthogonally to the treatment head and its EPID imaging device. The kV system shares a common axis of rotation with the MV treatment source. The kV beam is 425×425 mm² incident on the flat-panel detector. Images can be acquired with three different fields-of-view (FOV); small, medium, and large.

The difference between the three is the edge to the kV central beam, which is 138 mm for the small FOV, 213 mm for the medium FOV, and 262 mm for the large field of view. During 360 deg rotations the system acquires approximately 650 projection images. The size of the flat-panel detector is 410×410 mm² and is located at a fixed source to detector distance, resulting in limited FOVs along the x and z direction. An offset scanning geometry was used, in which the imager was shifted laterally 10 cm and a corresponding asymmetric beam, defined by an M20 collimator, was utilized to scan a larger FOV. This offset geometry [Fig. 1(b)] was used to provide the optimum image sets, in terms of quality and reconstructed FOV. The Feldkamp algorithm for 3D filtered

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**Fig. 1.** (a) A photograph of the cone-beam CT Elekta Synergy CBCT system. (b) A schematic of the CBCT offset geometry at angle 0 deg and medium FOV. (c) Schematic of the CBCT offset geometry and medium FOV after 180 deg rotation.

**Fig. 2.** (a) Photograph of the Elekta designed bowtie filter. (b) Bowtie thickness measured with needle gauge as a function of length.
back-projection algorithm was used for CBCT reconstruction, with a Hann reconstruction filter. The total reconstructed volume used in this work was $410 \times 410 \times 264$ mm$^3$ with voxel size of $1 \times 1 \times 1$ mm$^3$.

II.B. Bowtie filter

An aluminum bowtie filter (Elekta, Filter Cassette Assembly, F1) was employed for all the tests described here. This F1 filter was inserted in the filter tray 30 cm from the x-ray source on an Elekta Synergy XVI unit. For the medium FOV, a cassette of M20 collimator was inserted between the x-ray tube and F1 filter shown in Fig. 2(a). There are various possible designs for bowtie filters implemented on Elekta Synergy XVI systems. Bowtie filters could potentially be optimized based on a variety of imaging tasks while accounting for tradeoffs between image quality and patient dose. The design of the bowtie filter used here is based on patient size, dose, nominal x-ray energy, and material.

The bowtie filter used in this study [Fig. 2(a)] has no modulation in the z direction. The filter used for this work was built based on the objective of achieving uniform fluence through a cylindrical water phantom of diameter 300 mm measured at 120 kVp. The manufacturing material was aluminum of size $85 \times 135$ mm$^2$. The bowtie filter thickness measured with a needle gauge as a function of length is shown in Fig. 2(b). The filter was rigidly mounted in the cassette with a 1 mm thick acrylic covering protecting it from physical damage or scratching.

II.C. Dosimetry measurements

Dose measurements were made with and without a bowtie filter, such that the scanning geometry, phantom size, beam quality, and mAs/projections were the same for both cases. The phantom (shown in Fig. 3[a]) was a PMMA CTDI dosimetry phantom with diameter of 320 mm and thickness of 150 mm (RTI Electronics, Mölndal, Sweden). Two low-density polyethylene (LDPE) blocks ($32 \times 32 \times 15$ cm$^3$) were placed superior and inferior to the acrylic phantom to simulate scattered dose. The center of the phantom was placed at the machine isocenter position. This specially designed phantom allows for accurate positioning of radiation detectors. A 0.6 cc farmer-type ion chamber (NE 2571, Nuclear Associates) and Fluke 3504 electrometer at atmospheric pressure 756.2 mmHg and room temperature $22 \degree C$ was used for dose measurements. The dose was measured at five different radial positions of the phantom’s center to periphery. The ratio of doses with and without bowtie was modeled for all the radial positions of the phantom’s center to periphery. This model contained four fitting coefficients and is very useful to compute the dose reduction attributed to the bowtie at any radial position of the CTDI phantom.

II.D. Scatter-to-primary ratio measurements

Scatter-to-primary ratio was measured for imaging conditions with and without the bowtie filter using a beam-blocker method.\textsuperscript{9} The SPR can be defined as

$$\text{SPR} = \frac{S}{P},$$

where $S$ is the energy integrated signal of the scattered radiation measured as the average of a $10 \times 10$ pixel$^2$ area on
the FPI, and $P$ is the signal from the primary radiation. The same CBCT system/geometry was used as discussed in Sec. II A. The SPR measurements were performed at the center and edge of the projection image for the circular and irregular [posterior-anterior (PA) view] phantoms. A $10 \times 10 \times 4 \text{ mm}^3$ lead block was placed on the central axis before the phantom on the side facing the source. The scatter accounting for shadowing a part of the phantom by the block was not included for this SPR data. Twenty (20) images of the circular and irregular phantoms were acquired with and without the bowtie filter at the tube voltage of 120 kVp and exposures 1.0 and 1.6 mAs, respectively. The SPR measurements were performed on an irregular phantom at equivalent patient dose techniques. Measurements were performed on an irregular phantom at equivalent exposures 1.0 and 1.6 mAs, respectively. The SPR measurements were performed on an irregular phantom at equivalent patient dose techniques (0.1 mGy/exposure) for with and without the bowtie filter. This required an exposure of 1.6 mAs/projection (bowtie filter) as compared to 1.0 mAs/projection (without bowtie filter) at 120 kVp. The same procedure was used for the SPR measurements at the phantom’s edge except the block was moved from the center to the edge of the phantom’s projection.

II.E. CBCT image quality: Phantom study

The Catphan 500 (The Phantom Laboratory, Salem, NY) was employed in these investigations. Experiments were performed with the Catphan 500 phantom alone (referred to as “circular phantom”) and with the Catphan 500 phantom inserted in an irregular annulus (referred to as “irregular phantom”; Fig. 3(e)] shaped to reflect a humanoid torso. Two types of configurations were used to evaluate overall CBCT image quality with and without a bowtie filter. The phantom was positioned at the center of the imaging field of view with the help of room lasers. A photograph of the cone-beam CT system is shown in Fig. 1(a). Image acquisition proceeded with gantry rotation over 360 deg for all the cases during which approximately 660 planar images were acquired at the medium FOV. Images of the circular phantom, with and without bowtie filter, were acquired under the same imaging conditions at 120 kVp and tube current 40 mA and exposure time 20 ms per projection. Similarly, images of the irregular phantom with and without bowtie filter were acquired under the same imaging conditions at 120 kVp and tube current 40 mA and exposure time 40 ms, respectively. The CBCT reconstruction algorithm includes a scatter correction technique as described by Boellaard et al. The irregular phantom was scanned using a conventional helical CT scanner (Discovery ST 16 slice, GE Healthcare, Milwaukee, WI) at 120 kVp and 300 mAs. The total number of CT reconstructed slices were 264 with each slice thickness of 1 mm and voxel size of 1 $\times 1 \times 1 \text{ mm}^3$. The CT number for both CBCT and CT images was defined as $\text{CT}_{a} = [(\mu_{\text{object}} - \mu_{\text{water}})/\mu_{\text{water}}] \times 1000$ HU, where $\mu_{\text{object}}$ and $\mu_{\text{water}}$ are the linear attenuation coefficients for the object and water, respectively. Quantitative comparison of image quality based on skinline recovery, CT number accuracy, CT number uniformity, spatial resolution, and contrast-to-noise ratio were evaluated using the circular phantom tests performed in acquisitions with and without the bowtie filter. Imaging metrics were also performed on images generated on a conventional CT scanner.

II.E.1. Uniformity

The uniformity measurements were performed on the circular and irregular phantoms. The circular phantom contains a uniform, water-equivalent portion (CTP486) with CT number (HU) within −10 and +10 HU at standard scanning protocols. The CBCT images were analyzed in Matlab. Five ROIs of size $10 \times 10 \text{ mm}^2$ were selected in CBCT images of the phantom—one at the center and four at peripheral positions symmetrically around the center, each within the central axial plane. The spatial nonuniformity (SNU) was defined as

$$\text{SNU} = \frac{V_{\text{max}} - V_{\text{min}}}{1000} \times 100, \quad (2)$$

where $V_{\text{max}}$ and $V_{\text{min}}$ are the maximum and minimum mean voxel values, respectively, within each ROI.

II.E.2. Skinline artifacts/Reduction in CT number at skin zone

In CBCT images, the reduction in CT number (RCTN) at the skin zone is attributed to the detector saturation near the object periphery. Measurements were performed on circular and irregular phantoms. The reduction in CT number (denoted $\Delta\text{CT}_{a}$) was measured using 3 ROIs at skin depths 5, 10, and 20 mm. The size of each ROI was $4 \times 4 \text{ mm}^2$. The average reduction in CT (RCTN) was calculated as

$$\text{RCTN} = \frac{\overline{\text{HU}_{\text{CT}}}}{\overline{\text{HU}_{\text{CBCT}}}} - 100, \quad (3)$$

where $\overline{\text{HU}_{\text{CT}}}$ and $\overline{\text{HU}_{\text{CBCT}}}$ are the mean voxel values for a given ROI measured in helical CT and CBCT images, respectively.

II.E.3. CT number accuracy and linearity

The CT number accuracy and linearity measurements were performed on circular and irregular phantoms. The phantom contains seven different cylindrical contrast inserts (12 mm diameter, CTP 404) with known electron densities and CT numbers. The mean HU value was measured for each contrast insert. The CBCT images were converted into polar coordinates for convenient analysis. A 2D ROI size was approximately $4 \times 4 \text{ mm}^2$ covering approximately 16 voxels and was confined to within the contrast inserts. For conventional CT images, the 2D ROI size was $4 \times 4 \text{ mm}^2$. The voxel values in helical CT images of the phantom defined the true CT number.

II.E.4. Contrast-to-noise ratio

CNR measurements were performed using the circular and irregular phantoms (CTP 404 module), including polystyrene (PS, −100 HU) and low-density polyethylene (LDPE, −35 HU) inserts of 12 mm diameter. Contrast-to-noise ratio (CNR) measurements were performed on circular
and irregular phantoms. For each individual phantom, the imaging conditions including kVp, exposure time, and tube current were the same, with and without the bowtie filter. ROIs of size 4×4 mm² (~4×4 voxels) within each insert were used to measure the mean and standard deviation (noise) in HU value. The mean HU value was measured for each contrast insert. The CNR (signal difference divided by average noise) values was calculated as

\[
\text{CNR} = 2 \times \frac{\overline{\text{HU}(\text{LDPE})} - \overline{\text{HU}(\text{PS})}}{\text{Noise}(\text{LDPE}) + \text{Noise}(\text{PS})},
\]

where \(\overline{\text{HU}(\text{LDPE})}\) and \(\overline{\text{HU}(\text{PS})}\) are the mean voxel values in LDPE and PS, respectively, and \(\text{Noise}(\text{LDPE})\) and \(\text{Noise}(\text{PS})\) are the standard deviation in voxel values in LDPE and PS, respectively.

### II.E.5. Spatial resolution and modulation transfer function

The MTF measurements were performed on circular and irregular phantoms using the CTP528 module, which contains a radial high contrast spatial resolution bar pattern ranging from 1 through 21 line pair per cm. A bar pattern may be considered resolved if the bars can be perceived with some discernible spacing or lowering density among them. The radial design eliminates the possibility of streaking artifacts from other test objects. A circular pattern of pixels (passing through the bar patterns) was taken to obtain square-wave response function (SWRF). The amplitude response at various spatial frequencies was analyzed between 1 and 21 lp/cm. The MTF of the system was determined by deconvolving the SWRF.¹⁵

### II.F. Patient study

CBCT image quality measurements including uniformity and skinline reconstruction were performed on ten gynecological patients. But only one patient is reported here because there were no significant differences from patient to patient. Images were acquired with and without a bowtie filter at 120 kVp and 1.6 mAs (~650 projections acquired across 360 deg). The same CBCT system geometry and procedure was used as explained in the phantom study section. CBCT reconstructions with and without bowtie filter were analyzed quantitatively (difference images) and qualitatively (examination by a radiation oncologist). Of specific interest in these studies were the gynecological patient contrast and the reduction in CT number (cupping artifact) from the skinline to the center of the image.

### III. RESULTS AND DISCUSSIONS

#### III.A. Dose measurements for kV CBCT

The uncertainty of the ion chamber measurements was determined (standard deviation/mean value) to be within ±0.55%. The dose reduction (define in methods) with bowtie filter as a function of radial distance from the phantom center is shown in Fig. 3(b). It shows the achievable decrease in dose at the center and periphery of the body phantom with the bowtie used in this study. Dose reductions by 25% and 43% were found at phantom center and skin depth of 1 cm, respectively. It may be noted that the increase in dose reduction was consistent and followed a Lorentzian peak function (LPF). A theoretical fit represented by a solid line is shown in Fig. 3(b). Dose reduction with a bowtie as a

### Table I. The scatter, primary, and scatter-to-primary ratios behind the center and edge of the circular (Cat-Phan500) phantom using lead strip of size 10×10×3.5 mm³.

<table>
<thead>
<tr>
<th>Scanning techniques</th>
<th>Scatter at phantom center (ADU)</th>
<th>Primary at phantom center (ADU)</th>
<th>Scatter at phantom periphery (ADU)</th>
<th>Primary at phantom periphery (ADU)</th>
<th>Scatter-to-primary ratio at center</th>
<th>Scatter-to-primary ratio at edge</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bowtie 120 kVp/0.8 mAs</td>
<td>302</td>
<td>801</td>
<td>602</td>
<td>6594</td>
<td>0.377 ± 0.02</td>
<td>0.092 ± 0.02</td>
</tr>
<tr>
<td>Without bowtie 120 kVp/1.6 mAs</td>
<td>630</td>
<td>1212</td>
<td>1301</td>
<td>12730</td>
<td>0.519 ± 0.03</td>
<td>0.102 ± 0.02</td>
</tr>
</tbody>
</table>

### Table II. The scatter, primary, and scatter to primary ratios behind the center and edge of the irregular (Cat-Irreg) phantom using lead strip of size 10×10×3.5 mm³.

<table>
<thead>
<tr>
<th>Scanning techniques</th>
<th>Scatter at phantom center (ADU)</th>
<th>Primary at phantom center (ADU)</th>
<th>Scatter at phantom periphery (ADU)</th>
<th>Primary at phantom periphery (ADU)</th>
<th>Scatter-to-primary ratio at center</th>
<th>Scatter-to-primary ratio at edge</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bowtie 120 kVp/1.6 mAs</td>
<td>1011</td>
<td>2280</td>
<td>2539</td>
<td>20427</td>
<td>0.443 ± 0.03</td>
<td>0.124 ± 0.01</td>
</tr>
<tr>
<td>Without bowtie 120 kVp/1.6 mAs</td>
<td>2125</td>
<td>3221</td>
<td>4240</td>
<td>31921</td>
<td>0.661 ± 0.03</td>
<td>0.133 ± 0.01</td>
</tr>
<tr>
<td>Without bowtie 120 kVp/1.0 mAs</td>
<td>1328</td>
<td>2059</td>
<td>2625</td>
<td>1937</td>
<td>0.645 ± 0.025</td>
<td>0.132 ± 0.01</td>
</tr>
</tbody>
</table>
The SPR at the center of each phantom is higher than the SPR measured at the edges. But the absolute scatter signal is higher at the edges of both phantoms including circular and irregular. The SPR values for the circular and irregular phantoms for with and without bowtie filter are listed in Tables I and II, respectively. It shows that the scatter signal at the phantom’s edge is higher than the primary signal at the center for the case without the bowtie filter. The absolute scatter signal at the edges of both phantoms is almost two times less than without the bowtie filter. This high absolute scatter signal strongly affects the SPR behind the center of the phantom and hence image quality. Also, the SPR values at the center and edge are higher for the irregular phantom because of the size and excessive fluence at the edge that creates high scatter and affects the SPR at the center (throughout the image). The bowtie filter shapes and modulates the beam to reduce the unwanted signal at the object’s periphery and hence improves the SPR behind the phantom’s center and prevents detector saturation at the skin zone.

III.C. CBCT image quality: Phantom study

III.C.1. Uniformity and skinline

Trans-axial CBCT images of a uniform water-equivalent portion of the circular phantom acquired without and with bowtie filter are shown in Figs. 4(a) and 4(b), respectively. Profiles through the images are shown in Fig. 4(c); the profile position is indicated by a dotted line in the image. Image [Fig. 4(a)] and its profile both show severe reduction in CT number near the skin zone, which reduces as a function of skin depth. This is due to the signal range of the detector (saturation) at the periphery of the phantom, leading to a loss of information in projections that creates a skinline artifact in the reconstruction. Approximately 15% of average reduction in CT number was measured at 1 cm skin depth. The CBCT image with the bowtie filter [Fig. 4(b)] and its profile [Fig. 4(c)] demonstrate significant improvement in the recovery of CT number at the skin zone compared to the nominal case. Almost 100% of the CT number is restored at 1 cm skin depth. The bowtie filter improves image uniformity by (i) flattening of the fluence profile at the detector (thereby reducing the dynamic range requirement and preventing detector saturation at the skinline) and (ii) attenuates more fluence at the edge of the field which reduces the primary beam on the periphery of the phantom, which, in turn reduces scatter throughout the image. The reduction in scatter throughout the image has the greatest advantage in the center of the image where the bowtie has minimally affected the primary fluence. The mean spatial non-uniformity, as defined in Eq. (2), was 2.1% and 9.8% for image acquired with and without a bowtie filter, respectively. The higher non-uniformity number as defined in Eq. (2) means poorer uniformity.

Trans-axial CBCT images of irregular phantom acquired without and with bowtie filter are shown in Figs. 5(a) and 5(b), respectively, at the same window/level. The difference image (image without bowtie subtracted from the image acquired with bowtie) is shown in Fig. 5(c). Profiles through the reconstructed slices of CBCT images are compared with helical CT in Fig. 5(d), with the bowtie filter case providing an improvement in edge definition. The images and profiles show a severe reduction of CT number near the skin zone in the nominal CBCT image. Since the irregular phantom is larger in size than the circular phantom, a higher exposure is required for sufficient signal behind the center of the object, resulting in increased pixel saturation near the phantom periphery, which is the main source of CT number reduction at the skinline. Even with the bowtie filter case, there is still some missing skin on the right shoulder of the image profile in Fig. 5(d). This is due to the detector lag and ghosting effects. As the object is irregularly shaped, the projected location of skinline relative to previous projections causes an accumulated lag signal that degrades the reconstruction and results in reduction in CT number in the skinline zone. The gantry rotates clockwise around an irregular phantom, which...
FIG. 5. Trans-axial CBCT images of irregular (Cat-Irreg) phantom (a) without (window level: $-350$ to $500$ HU) and (b) with bowtie filter (window level: $-350$ to $500$ HU). (c) The difference of image (b) and (a). (d) Profiles through the uniform water portion of the irregular phantom. (e) The right shoulder of (d) is magnified to see the missing skinline more clearly. (f) Reduction in CT number (RCTN) at several skin depths for with and without bowtie filter.

FIG. 6. Trans-axial images of circular phantom with several contrast inserts acquired (a) without (window level $-800$ to $500$ HU) and (b) with bowtie filter (window level $-800$ to $500$ HU). (c) The same image shown in (a), but transformed into polar coordinate (window level $-800$ to $500$ HU). (d) Profiles through the image (c) with several inserts are compared with CT and without bowtie filter. (e) The difference between the measured CT and CBCT number plotted as a function of Ideal CT number for several inserts. The empty circle and filled square symbols represent CT number error with and without bowtie. A gray solid line represents a linear fit to bowtie data.
causes lag on the right side of the image. Mail et al. quantitatively explained this lag effect in CBCT images of irregular phantom acquired with and without bowtie filter. Even without a bowtie filter, the reduction in CT number at the skinline is worse on the right shoulder than the left in Fig. 5(d), due to the lag and ghosting. This clearly explains the bowtie case where the reduction in CT number at the skinline on the right shoulder in Fig. 5(d) is due to lag and ghosting (not due to the compatibility of the bowtie filter for an irregular shaped phantom). The shoulder on the right hand side of Fig. 5(d) is magnified in Fig. 5(e) to illustrate the reduced signal at the periphery. The mean reduction in the CT number at the skin region, as defined in Eq. (3), is quantitatively shown in Fig. 5(f) for three different skin depths for both cases. The mean signal losses for skin depths of 5, 10, and 20 mm were found 35.1%, 25.15%, and 20.12%, respectively, in cases without bowtie filter images. These numbers in bowtie cases were measured 13.8%, 12.5%, and 11%.

The spatial non-uniformities of the irregular phantom without and with bowtie filter [Figs. 5(a) and 5(b)] were calculated to be 4.1% and 5.7%, respectively. Profiles in Fig. 5(d) show that the CT and CBCT images with the bowtie filter have similar uniformity within the circular phantom region.

### III.C.2. CT number accuracy and linearity

The central slice images of a circular phantom without and with bowtie filter are shown in Figs. 6(a) and 6(b), respectively. For convenient analysis, the image in Fig. 6(b) was transformed into polar coordinate in Fig. 6(c). A profile through several inserts, indicated by a dotted line in image, is shown in Fig. 6(d). The dotted, solid black, and gray solid lines represent CT and CBCT with and without bowtie filter image profiles, respectively. This signal profile shows that CBCT image acquired with a bowtie filter is comparable to helical CT. The comparison of CT number accuracy and linearity for circular phantom, without and with the bowtie filter, is shown in Fig. 6(e), where the difference of measured CT number to CBCT is plotted versus the ideal CT number. The CBCT with bowtie filter demonstrated a clear improvement in CT number accuracy and an increase in image contrast compared to the nominal cases. For the nominal case, inaccuracy in CT number is related to the assumption that the detected signal is all primary fluence. Scattered fluence clearly violates this assumption. Reducing scatter will lead to more accurate CT numbers by allowing accurate inversion of

<table>
<thead>
<tr>
<th>Techniques</th>
<th>Slope</th>
<th>R²</th>
<th>Intercept</th>
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</thead>
<tbody>
<tr>
<td>CT</td>
<td>0.98</td>
<td>0.995</td>
<td>4</td>
</tr>
<tr>
<td>Bowtie</td>
<td>0.85</td>
<td>0.98</td>
<td>43</td>
</tr>
<tr>
<td>Without bowtie</td>
<td>0.71</td>
<td>0.98</td>
<td>-3</td>
</tr>
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Fig. 7. Trans-axial CBCT images of irregular (Cat_Irreg) with several inserts, (a) without bowtie (window level –350 to 500 HU) and (b) with bowtie filter (window level –350 to 500 HU). (c) The difference of image (b) and (a). (d) The same image in (a) but converted into polar coordinates (window level –700 to 300 HU). (e) The same image in (b) but converted into polar coordinates to make it simple for analysis such as CT number accuracy and streaking artifacts (window level –700 to 300 HU). (f) The measured CT number plotted as a function of ideal CT number for several inserts.
Beer’s law. It should be noted that other invalid assumptions (e.g., beam hardening) can lead to inaccuracies in CT number as well. The improvement in CT number accuracy with the bowtie filter is primarily attributed to the scatter reduction (with improved beam-hardening and non-uniformity also contributing to improved CT number accuracy) throughout the image and particularly near the phantom’s periphery. This reduction in scatter reduces the scatter to primary ratio and hence improves CT number accuracy. The CT number accuracies with and without a bowtie filter were found to be ±21 and ±68 HU, respectively.

Central slice images of the irregular phantom with and without a bowtie filter are shown in Figs. 7(a) and 7(b), respectively. These images were then transformed into polar coordinates to make the image analysis simple. The transformed polar coordinate images are shown in Figs. 7(d) and 7(e). In Fig. 7(f), the measured CT number for Catphan phantom within irregular annulus is plotted versus the ideal CT number. The measurements performed with a bowtie filter showed greater linear CT number accuracy than nominal. The mean CT number obtained from a conventional CT image showed reasonably good agreement with the expected CT numbers. Detailed comparisons of the CT number accuracy and linearity in terms of slope, $R^2$ values, and intercepts are shown in Table III. The slope of the fitting line and intercept values are reasonably improved. The CT number accuracy for the helical CT and CBCT acquired with a bowtie as well. The improvement in CNR is significantly reduced for the irregular phantom than for the circular phantom, while this reduction is very small in the case of the bowtie filter. This is because of the scatter compensation. Comparing images acquired at equivalent patient dose (~0.1 mGy/exposure), the comparison of CNR improves even further, giving a CNR of 4.22 ± 0.41 for the bowtie, compared to 1.61 ± 0.3 without a bowtie.

The central slices of irregular phantom converted into polar coordinates with and without the bowtie filter are shown in Figs. 7(e) and 7(d), respectively. The streaking artifacts are more visible around the high-density inserts in the polar coordinate image acquired without the bowtie filter. These streaking artifacts appear due to scatter and beam hardening. Beam hardening and scatter artifacts are worse because of the large size and irregular shape of the phantom and the presence of high contrast objects. The image in Fig. 7(e) acquired with a bowtie filter is almost free of streaking artifacts. The scatter compensation reduces streaking artifacts and improves CNR. The bowtie filter reduces beam hardening effects by minimizing the range of variation in x-ray energies presented to the detector. For example, considering the PA projection of the irregular phantom, the mean energy of the x-ray beam is 75 keV at the center of the detector (path length 320 mm acrylic) and 64 keV near the periphery (path length 100 mm acrylic). With a bowtie filter, however, the range in energy is reduced to 76 keV at the center and 76 keV at the periphery. Beam-hardening artifacts are therefore expected to be mitigated significantly by the bowtie filter.

### III.C.3. Contrast-to-noise ratio and streak artifacts

CNR measurements were performed on Catphan and irregular phantoms (contrast inserts, LDPE and Polystyrene). For Catphan phantom, the CNR, as defined in Eq. (4), was measured as 4.51 ± 0.43 and 3.58 ± 0.35, with and without a bowtie filter, respectively (an improvement by a factor of 1.3 at fixed mAs). The improvement is likely due primarily to reduced scatter (with improved beam-hardening and non-uniformity also contributing to improved CNR). The CNR values for the irregular phantom were found to be 4.22 ± 0.41 and 2.24 ± 0.32, with and without bowtie cases, respectively (at equivalent mAs). The improvement in CNR is found to be much greater for the irregular than circular phantom. A primarily possible approach is that for the without bowtie case, the amount of scatter in the irregular phantom is higher than in the circular phantom, particularly at the phantom’s peripheries. The CNR in the nominal case is significantly reduced for the irregular phantom than for the circular phantom, while this reduction is very small in the case of the bowtie filter. This is because of the scatter compensation. Comparing images acquired at equivalent patient dose (~0.1 mGy/exposure), the comparison of CNR improves even further, giving a CNR of 4.22 ± 0.41 for the bowtie, compared to 1.61 ± 0.3 without a bowtie.

The central slices of irregular phantom converted into polar coordinates with and without the bowtie filter are shown in Figs. 7(e) and 7(d), respectively. The streaking artifacts are more visible around the high-density inserts in the polar coordinate image acquired without the bowtie filter. These streaking artifacts appear due to scatter and beam hardening. Beam hardening and scatter effects are worse because of the large size and irregular shape of the phantom and the presence of high contrast objects. The image in Fig. 7(e) acquired with a bowtie filter is almost free of streaking artifacts. The scatter compensation reduces streaking artifacts and improves CNR. The bowtie filter reduces beam hardening effects by minimizing the range of variation in x-ray energies presented to the detector. For example, considering the PA projection of the irregular phantom, the mean energy of the x-ray beam is 75 keV at the center of the detector (path length 320 mm acrylic) and 64 keV near the periphery (path length 100 mm acrylic). With a bowtie filter, however, the range in energy is reduced to 76 keV at the center and 76 keV at the periphery. Beam-hardening artifacts are therefore expected to be mitigated significantly by the bowtie filter.

### III.C.4. Spatial resolution and modulation transfer function

The spatial resolutions (for 50% MTF) of the circular phantom’s images acquired with and without a bowtie filter were measured as 2.3 ± 0.11 and 2.2 ± 0.1 lp/cm, respectively. The MTF is essentially equivalent with or without a bowtie filter, with the curves of Fig. 8 showing consistent shape between the two cases. For the irregular phantom, the spatial resolutions (for 50% MTF) were measured as 2.1 ± 0.08 and 2.0 ± 0.06 lp/cm with and without a bowtie, respectively. The measured MTFs for with and without the bowtie filter have the same values within the error bar.
III.D. Patient study

CBCT images of a gynecological patient acquired without and with a bowtie filter are shown with the same window level in Figs. 9(a) and 9(b), respectively. Profiles through the images are shown in Fig. 9(c). Moderate improvements in uniformity and skinline reconstruction are seen with the bowtie filter. Improvement in skinline reconstruction is clearly shown in Fig. 9(d), which is the magnified portion indicated on the right of Fig. 9(c). The black spots in the images are a brachytherapy applicator used as a fiducial marker. The image in Fig. 9(a) demonstrates less clarity due to high scatter-to-primary ratio while the image in Fig. 9(b) demonstrates clarity due to reduction in scatter-to-primary with the bowtie filter.

IV. DISCUSSIONS

A fixed, shaped filter has been used to selectively attenuate x rays at the periphery of the exposed field-of-view to compensate for the patient profile. The filter reduces the dynamic range demands on the detector during a CBCT scan. The resulting flux on the detector is relatively uniform across the detector and does not result in saturation in regions beyond the skinline. Further, the reduction of the excessive fluence near the skin zone with a bowtie filter reduces the absolute scatter signal at the object’s edge. This reduction in the absolute scatter signal improves the SPR behind the center of the phantom and hence image quality. The influence of the bowtie filter on both dose and image quality was evaluated. The application of the bowtie filter for circular and irregular phantoms resulted in an improvement in image quality and reduction/alteration in the dose distribution within the patient. Figure 3(b) demonstrates that the application of the bowtie filter results in a reduction in dose by 43% at the skinline and 26% at the phantom center. Assessment of contrast-to-noise performance demonstrated an improvement when the bowtie filter is employed. This was specifically observed in the large sized phantom (irregular phantom) where scatter is increased. Overall, the bowtie improves CT number accuracy and image uniformity. This improvement is attributed to the scatter reduction, beam hardening reduction, and flatness of fluence at the detector.

The selection/design of a bowtie filter is challenging and will always be less than ideal. It is not likely to be optimum given that a subject being imaged is significantly less than uniform, not exactly cylindrical in shape, and not necessarily centrally located in the x-ray beam. In such cases, it is possible for one or more disjointed regions of saturation to occur or conversely to over-filter the x-ray flux and unnecessarily create regions of very low signal. Regardless of compromises associated with patient size, shape, anotomical site, and field of view, the use of a simple bowtie filter results in an overall improvement in the quality of CBCT images.

V. CONCLUSION

The implemented bowtie filter shows reduction in scatter-to-primary ratio for the circular and irregular phantoms. The bowtie filter shows an improvement in image quality including uniformity, CT number accuracy, contrast-to-noise ratio, and skinline reconstruction of the circular and irregular phantoms. The implementation of a bowtie filter also reduced streaking artifacts in CBCT images of an irregular phantom compared to nominal. The implemented bowtie filter demonstrated an improvement in skinline reconstruction and uniformity in images of gynecological patients. This compensator
is static and makes many compromises for anatomical imaging site, patient size, and imaged field of view. The ideal compensator would optimize the fluence profile to account for numerous properties of the patient and imaging system. The investigated performance of the bowtie filter on image quality of CBCT will create confidence in clinical implementation of bowtie filters.

Author to whom correspondence should be addressed. Address for correspondence: Radiation Physics, Radiation Medicine Program, Princess Margaret Hospital, 610 University Avenue, Toronto, Ontario M5G 2M9, Canada. Telephone: 416-946-4501 X5384; Fax: 416-946-6566; Electronic mail: david.jaffray@rmp.uhn.on.ca