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Prior image constrained scatter correction in cone-beam computed tomography image-guided radiation therapy

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Abstract

X-ray scatter is a significant problem in cone-beam computed tomography when thicker objects and larger cone angles are used, as scattered radiation can lead to reduced contrast and CT number inaccuracy. Advances have been made in x-ray computed tomography (CT) by incorporating a high quality prior image into the image reconstruction process. In this paper, we extend this idea to correct scatter-induced shading artifacts in cone-beam CT image-guided radiation therapy. Specifically, this paper presents a new scatter correction algorithm which uses a prior image with low scatter artifacts to reduce shading artifacts in cone-beam CT images acquired under conditions of high scatter. The proposed correction algorithm begins with an empirical hypothesis that the target image can be written as a weighted summation of a series of basis images that are generated by raising the raw cone-beam projection data to different powers, and then, reconstructing using the standard filtered backprojection algorithm. The weight for each basis image is calculated by minimizing the difference between the target image and the prior image. The performance of the scatter correction algorithm is qualitatively and quantitatively evaluated through phantom studies using a Varian 2100 EX System with an on-board imager. Results show that the proposed scatter correction algorithm using a prior image with low scatter artifacts can substantially mitigate scatter-induced shading artifacts in both full-fan and half-fan modes.

1. Introduction

Cone-beam CT imaging has become an area of active research due to its ability to accomplish volumetric reconstructions in a single gantry rotation with comparable image quality relative to conventional fan-beam CT (Jaffray and Siewerdsen 2000). As a result, cone-beam CT has shown a variety of promising applications. Namely cone-beam CT imaging can be performed prior to radiation therapy treatment to verify patient setup (Jaffray *et al* 2002, Oldham *et al* 2005). In addition, cone-beam CT can be used in other image-guided radiation therapy applications, such as brachytherapy (Al-Halabi *et al* 2010, Westendorp *et al* 2007) and intraoperative procedures (Khoury *et al* 2007a, 2007b, Rafferty *et al* 2006), and can serve as a powerful image-guidance tool when incorporated in C-arm technology (Siewerdsen *et al* 2005, Wallace *et al* 2009, Rafferty *et al* 2005, Daly *et al* 2008, Orth *et al* 2008, Knight *et al* 2008, Zambelli *et al* 2008).

However, it has been demonstrated that scatter becomes significant when thicker objects and larger cone angles are used, resulting in reduced contrast and CT number inaccuracy (Siewerdsen and Jaffray 2001). To combat the effects of scatter, a variety of methods have

been proposed. Initial inquiries were aimed at the optimization of imaging geometry (Siewerdsen and Jaffray 2000), the incorporation of anti-scatter grids (Siewerdsen *et al* 2004, Kyriakou and Kalender 2007), and the use of bowtie filters (Mail *et al* 2009). Other methods have used analytical approaches to model scatter (Zbijewski and Beekman 2006, Star-Lack *et al* 2009, Spies *et al* 2000, 2001, Boone and Seibert 1988) or used Monte Carlo simulations to model scatter distributions (Jarry *et al* 2006, Kyriakou *et al* 2006, Mainegra-Hing and Kawrakow 2008, Chan and Doi 1983, Chan and Doi 1985, Poludniowski *et al* 2009), while others have pursued experimental techniques to measure or estimate the contribution of scatter in projections (Siewerdsen *et al* 2006, Ning *et al* 2004, Zhu *et al* 2009a). Most recently, Zhu *et al* (2009b) proposed measuring the scatter distribution at the initial day of treatment with a partially blocked cone-beam CT scan, subsequently applying corrections to later cone-beam CT scans of the same patient.

Advances have been made in x-ray computed tomography (CT) by incorporating a high quality prior image into the image reconstruction process (Chen *et al* 2008a, 2008b, 2009). It turns out that many of the characteristics of a high quality prior image can be imparted to the target image provided that the prior image is appropriately incorporated into the image reconstruction process. In this paper, we extend this idea to correct scatter-induced shading artifacts in cone-beam CT image-guided radiation therapy. Namely, we ask: Can we take advantage of prior images with low scatter artifacts to reduce shading artifacts in cone-beam CT images acquired under conditions of high scatter?

Recently, an empirical cupping artifact correction method was introduced and used to correct beam-hardening and bone-induced artifacts in micro-CT imaging (Kachelriess *et al* 2006). In this method, the to-be-reconstructed target image is written as a weighted summation of basis images, which are reconstructed using power-expansion of the raw cone-beam CT projection data. To determine the weight for each basis image, a homogeneous water phantom is scanned. Weighting coefficients of the basis images are found by fitting the basis images to a binary template, which has density ρ_w inside of the phantom and zero outside, and a weighting template, which excludes phantom walls and blurring due to point spread effects. Once the calibration is performed on a water phantom, the resulting coefficient vector is applied to projection data in subsequent small animal scans (Kachelriess *et al* 2006) with the same scanning parameters. A hybrid cupping correction method was also proposed (Sourbelle *et al* 2005), utilizing knowledge of the x-ray spectrum, as well as a dual energy application of the calibration technique using a micro-CT scanner (Stenner *et al* 2007). Although the results were demonstrated to be promising, a potential concern, with regard to scatter correction, is the feasibility of using the same expansion coefficients obtained from a homogeneous phantom scan to correct images for other image objects of varying size such as *in vivo* studies in image-guided procedures.

In this paper, instead of using a homogeneous phantom scan to determine the expansion coefficients, we use a prior image with low scatter artifacts of the same image object as a constraint to determine the expansion coefficients. The coefficients are then used in the expansion to generate the target image. Thus, the scanning of a water phantom is not needed for the determination of coefficients. In practice, this prior image may originate from a planning CT acquisition, which is acquired under conditions of low scatter. Due to the coordinate difference between the treatment planning CT scan and the cone-beam CT scan prior to the treatment delivery, a geometrical registration may be needed. In this study, for simplicity, a full scan with narrow *z*-coverage is used to simulate an image with low scatter artifacts, while a full scan with wide x-ray collimation is used to simulate a typical, clinical cone-beam CT scan, which often contains large contributions of x-ray scatter. Due to the fact that the extra radiation dose to generate the narrow *z*-coverage CT image is low, the method could be potentially used in practice.

The remainder of this paper has the following structure. In section 2, the basic principles of the prior image-based scatter correction algorithm are described. In section 3, results are provided for phantoms studies in the full-fan and half-fan mode. Section 4 provides a discussion of the performance of the scatter correction algorithm.

2. Methods and materials

2.1. Prior image constrained scatter correction

The correction algorithm requires two CT image volumes: a prior image volume with low scatter artifacts, X_{prior} , and a cone-beam CT image volume acquired under conditions of high scatter, X_{cone} , with projection data P_{cone} . The algorithm begins with a power-series expansion of the cone-beam projection data acquired under conditions of high scatter given by equation (1):

$$P' = \sum_{j=0}^M a_j P_{\text{cone}}^j \quad (1)$$

where P_{cone}^j represents cone-beam projection data raised to the j th power, a_j are the coefficients of the series and M is a constant. Then, filtered back projection is performed on P' , yielding image matrix X' :

$$X' = \sum_{j=0}^M a_j \text{FBP}(P_{\text{cone}}^j) = a_0 \text{FBP}(P_{\text{cone}}^0) + a_1 \text{FBP}(P_{\text{cone}}^1) + \cdots + a_M \text{FBP}(P_{\text{cone}}^M). \quad (2)$$

Thus, the correction algorithm is obtained by solving the following minimization problem:

$$\min \chi^2 = \min \|X' - X_{\text{prior}}\|_{l_2} \quad (3)$$

where $\|z\|_{l_2} = \sqrt{\sum_{i=1}^N |z_i|^2}$ is the l_2 norm of an N -dimensional vector $\mathbf{z}$. Equation (3) can be implemented via a least-squares method; that is, the coefficients of the series, a_j , can be found by taking partial derivatives of χ^2 with respect to a_j and setting them equal to zero (Bevington and Robinson 2003). In other words, the goal of the correction algorithm is to minimize the l_2 norm of the difference image between the prior image with low scatter artifacts and the filtered backprojection images of the power-series-expanded cone-beam projection data. Once the coefficients a_j are determined, they can be applied to their corresponding basis volumes and summed to achieve corrected image matrix X' . A flowchart of the algorithm is shown in figure 1.

2.2. Phantom evaluation of scatter correction algorithm in full-fan and half-fan modes

The correction algorithm is evaluated qualitatively and quantitatively using a Varian 2100 EX System equipped with an on-board imager. Three phantoms are used to test the performance of the correction algorithm in full-fan mode: the Catphan[®]500 Phantom, a Varian[®] water phantom, and the head portion of the Rando[®] phantom. Three phantoms are used to test the performance of the correction algorithm in half-fan mode: a CIRS Model 062 Electron Density Phantom, the shoulder portion of the Rando[®] phantom, and the pelvis portion of the Rando[®] Phantom. For each phantom, a 360° scan is acquired with narrow x-ray collimation in the z -direction to represent the image with low scatter artifacts. Then, a 360° scan is acquired with wide x-ray collimation in the z -direction such that the central slices of both acquisitions are identical; that is, the phantom remains stationary, while the x-

ray collimators are fully opened in the z -direction. Narrow and wide collimation scans are repeated using a full-fan bowtie filter in full-fan mode and a half-fan bowtie filter in half-fan mode. Table 1 shows relevant acquisition parameters. The source-to-center distance is 1000 mm, and the center-to-detector distance is 500 mm. The flat-panel detector has dimensions 397 mm $\times$ 292 mm with 1024 $\times$ 768 detector elements. Each 360° scan contains about 680 projections at angular spacing of 0.5°. In each configuration, images are reconstructed using a 512 $\times$ 512 matrix and the standard FDK algorithm (Feldkamp *et al* 1984). For half-fan geometries, the detector is laterally offset by 148 mm, and a pre-weighting scheme is used as described in Wang (2002).

3. Results

3.1. Evaluation of scatter correction algorithm in full-fan mode

Figure 2(a) shows the CBCT image of the Catphan 500 Phantom acquired in full-fan mode with narrow x-ray collimation without a full-fan bowtie filter. Figure 2(b) shows the reconstruction of the same slice from a subsequent acquisition with x-ray collimation fully opened. Figure 2(c) shows the application of the correction algorithm using the narrow collimation image as a prior image. No shading can be seen in the corrected image. To assess the CT number accuracy of the correction algorithm, regions of interest were placed in the contrast rods, as shown in figure 2(d). The mean values in HU for each ROI are evaluated. The percent differences were evaluated between the wide collimation and the prior image and the corrected image and the prior image, as seen in table 2, via the following equation:

$$\Delta = \frac{|HU_1 - HU_{\text{prior}}|}{HU_{\text{prior}}} \bullet 100\% \quad (4)$$

where Δ is the percent difference, HU_1 is the mean value of the region of interest and HU_{prior} is the mean value in the prior image. The same ROIs were evaluated with a full-fan bowtie, shown in table 3. The wide collimation image tends to deviate more from the prior image in the absence of a full-fan bowtie than in the presence of a full-fan bowtie, shown by the average 54% difference and average 33% difference, respectively. This indicates that the full-fan bowtie is able to reduce degradation due to scatter. When the correction algorithm is used, the method tends to be slightly more effective in the absence of a full-fan bowtie filter, as the corrected image retains an average 12% difference from the prior, while in the presence of a full-fan bowtie filter, an average 18% difference is observed. Note that the reduction in CT number in the interior of the phantom in the wide collimation image without a full-fan bowtie is severe with a 213% difference from the prior image, shown by ROI 9 in table 2. The correction method is able to reduce this CT number inaccuracy to 13% from the prior image. The same ROI, evaluated in the case of a full-fan bowtie, shows less degradation with only a 62% difference from its prior image, due to the influence of the full-fan bowtie. The correction method is able to improve the percent difference in this case to 22% from the prior image.

Figure 3 shows the CBCT images of the head portion of the Rando Phantom acquired in full-fan mode without a full-fan bowtie filter. Shading can be seen in the interior of the skull in figure 3(b). The corrected image in figure 3(c) has no shading. Figure 3(d) shows the placement of the line profile, which is displayed in figure 3(e). Reduction in CT number accuracy can be seen in the wide collimation profile. To better visualize the line profile, the central region is magnified and shown in figure 3(f). The restoration of reconstruction accuracy can be appreciated as the corrected profile more closely conforms to the prior image. Expansion coefficients, in the absence and presence of a full-fan bowtie filter, are presented in table 4.

3.2. Evaluation of scatter correction algorithm in half-fan mode

Figure 4(a) shows the CBCT image of the electron density phantom acquired in half-fan mode with narrow x-ray collimation without a half-fan bowtie filter. Figure 4(b) shows the reconstruction of the same slice from a subsequent acquisition with x-ray collimation fully opened. Figure 4(c) shows the application of the correction algorithm. Severe shading can be seen in figure 4(b), and visibility of low-contrast inserts is restored in the corrected image. To quantitatively evaluate CT number accuracy, ROIs are placed in the contrast rods of the electron density phantom, shown in figure 4(d). Mean values in HU of each ROI are presented in table 5. Percent differences in each ROI between wide collimation and prior image and between corrected image and prior image are evaluated via equation (4). As seen in table 5, the correction method is able to substantially reduce the average percent difference between wide collimation and prior image from 437% to an average percent difference of 88%, between the corrected and prior image.

Figure 5 shows the corresponding images when a half-fan bowtie filter is used with prior image shown in figure 5(a), wide collimation in (b) and corrected image in (c). Severe shading can be seen in figure 5(b), while the corrected image shows less shading and better visibility of the contrast rods. To quantitatively evaluate the CT number accuracy, figure 5(d) shows placement of ROIs. Table 6 shows the mean values in HU for each ROI. As was observed in the full-fan mode, the wide collimation image tends to deviate more from the prior image in the absence of a half-fan bowtie than in the presence of a half-fan bowtie, shown by the average 487% difference and average 389% difference, respectively. While this indicates that the half-fan bowtie filter is promoting CT number accuracy, an abnormal bright and dark pattern is visible in the interior of the phantom, seen in the wide collimation image of figure 5(b). Residual bright and dark spots can be seen in the corrected image in figure 5(c). The corresponding first, second and third power basis images, which contribute most to this corrected image, are shown in figure 6, revealing a similar pattern. Note that the first power basis image is the wide collimation image shown in figure 5(b) without conversion from linear attenuation coefficient (units of mm^{-1}) to HU. This abnormal artifact was observed in the pelvis phantom scans, shown in figure 7, specifically in the wide collimation image in figure 7(e), acquired with the half-fan bowtie. When the correction algorithm is used, the method tends to be slightly more effective in the absence of a half-fan bowtie filter, as the corrected image retains an average 88% difference from the prior, while in the presence of a half-fan bowtie filter, an average 105% difference is observed. The expansion coefficient values are summarized in table 7 in the absence and presence of a half-fan bowtie filter.

4. Discussion and conclusion

This study presents a new scatter correction method for cone-beam CT image-guided radiation therapy. The work is motivated by recent advances which have been made in x-ray computed tomography (CT) by incorporating a high quality prior image into the image reconstruction process (Chen *et al* 2008a, 2008b, 2009). Specifically, this study reveals that a scatter correction algorithm using a prior image with low scatter artifacts can substantially reduce scatter-induced shading artifacts in cone-beam CT images. The correction algorithm is evaluated through phantom studies using a Varian EX 2100 System with an on-board imager in the full-fan and half-fan modes.

This study reveals the differences in image quality that arise when objects of larger size are imaged in the half-fan mode, compared to the full-fan mode. In the full-fan mode, images of the Catphan phantom and Rando head phantom revealed traditional shading artifacts in the interior of each phantom. In comparison, in half-fan mode, the electron density and pelvis phantoms revealed a more abnormal artifact of bright and dark spots. This abnormal effect is

only slightly apparent without a half-fan bowtie in the electron density phantom image and is not visible in the pelvis phantom no-bowtie image, while the artifact is visible in electron density and pelvis phantom acquisitions with a half-fan bowtie. These observations have been reported in other studies (Star-Lack *et al* 2009, Zhu *et al* 2009b) and can be interpreted in light of the work of Virshup *et al* (2006). In their analysis, scatter profiles in the full-fan mode were found to be relatively flat and symmetric. Scatter profiles in the half-fan mode were found to be asymmetric across the detector with elevated scatter-to-primary ratios. The proposed method, due to the symmetry of the basis images, is well suited to counteract the traditional cupping artifact encountered in the full-fan mode; however, residual bright and dark spots can remain in corrected images in the half-fan mode, as asymmetries present in first-order basis images are further accentuated in higher order basis images and are challenging to counteract. The presence of these bright and dark distortions may be more appropriately treated with higher order or iterative methods.

There are several potential limitations in the presented study: (1) similar to other scatter measurements and correction methods (Star-Lack *et al* 2009, Ning *et al* 2004, Zhu *et al* 2009b), the proposed method may require a minimal amount of extra radiation to generate a prior image. This drawback may be overcome by directly using a treatment planning image as the prior image, although an image registration may be required to align the two image volumes and interfraction organ variation would need to be considered. (2) This study demonstrated a difference between a bowtie scatter rejection method and the proposed software correction method. However, this study also shares the same limitation as other published studies: there is no systematic study of performance comparison among different scatter artifacts correction methods. In future, more detailed comparative studies are needed to explore the extent to which the correction technique can be applied in clinical data.

There are several advantages and drawbacks to using prior image constrained scatter correction, with regard to the empirical cupping method, for scatter correction in cone-beam CT radiation therapy. The presented algorithm provides a potentially practical method for reducing scatter-induced shading artifacts as the correction algorithm only assumes that a prior image with low scatter artifacts is available. In comparison, if the empirical cupping method is to be used for this application, then it is beneficial to match the size and dimensions of the homogeneous phantom with those of the image object to avoid introducing additional image artifacts. A variety of homogeneous phantoms of various sizes and shapes would potentially need to be scanned using various x-ray tube settings for a given system to obtain the correct expansion coefficients, which may be a time-consuming process. On the other hand, the empirical cupping correction, because it uses a uniform template to constrain the calculation of expansion coefficients, is able to correct for beam hardening artifacts in the target image. In the proposed prior image constrained method, the target image can only perform as well as the prior image. Thus, if the prior image contains beam hardening artifacts, then when the target image is constrained to conform to it, beam hardening artifacts may still be present in the target image. Finally, in terms of dose, the proposed prior image constrained method may involve an additional narrow collimation scan. Thus, there may be an increase in dose with the proposed method, albeit, a minimal amount.

In conclusion, a prior image constrained scatter artifact correction method has been presented and validated in extensive phantom experiments in this paper. The method was demonstrated to substantially mitigate scatter-induced shading artifacts in phantom studies in both the full and half-fan modes used in image-guided radiation therapy procedures.

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Figure 1. Flowchart of the scatter correction algorithm. (This figure is in colour only in the electronic version)

Figure 2. Reconstruction of the Catphan[®]500 phantom without a full-fan bowtie filter. Display window: $[-100, 100]$ HU. FOV: $220 \text{ mm} \times 220 \text{ mm}$. (a) Prior image, (b) wide collimation image, (c) corrected image, (d) placement of ROIs.

Figure 3. Reconstruction of the head phantom without a full-fan bowtie filter. Display window: $[-100, 100]$ HU. FOV: $220 \text{ mm} \times 220 \text{ mm}$. (a) Prior image, (b) wide collimation image, (c) corrected image, (d) placement of line profile, (e) line profile, (f) zoomed in display of line profile.

Figure 4. Reconstruction of the electron density phantom without a half-fan bowtie filter. Display window: $[-300, 300]$ HU. FOV: $384 \text{ mm} \times 384 \text{ mm}$. (a) Prior image, (b) wide collimation image, (c) corrected image, (d) placement of ROIs.

Figure 5.

Reconstruction of the electron density phantom with a half-fan bowtie filter. Display window: $[-300, 300]$ HU. FOV: $384 \text{ mm} \times 384 \text{ mm}$. (a) Prior image, (b) wide collimation image, (c) corrected image, (d) placement of ROIs.

Figure 6.

Basis images for the electron density phantom with a half-fan bowtie. (a) First power, display window: [0.0086, 0.0123]; (b) second power, display window: [0.029, 0.069]; (c) third power, display window: [0.075, 0.39]. Note that the first power basis image is the wide collimation image presented in figure 5 without conversion from linear attenuation coefficient to HU. The gray scale window of each reconstructed image is $M \pm 2S$, where M and S are respectively the mean and standard deviation of an ROI covering each electron density phantom.

Figure 7.

Reconstructions of the pelvis phantom. Display window: $[-200, 200]$ HU. FOV: $414 \text{ mm} \times 414 \text{ mm}$. (a) No bowtie, prior image; (b) no bowtie, wide collimation image; (c) no bowtie, corrected image; (d) with half-fan bowtie, prior image; (e) with half-fan bowtie, wide collimation image; (f) with half-fan bowtie, corrected image.

Table 1

Acquisition parameters for scans in full-fan mode and half-fan mode.

	Full-fan mode		Half-fan mode	
	No bowtie	With full-fan bowtie	No bowtie	With half-fan bowtie
kVp	100	100	125	125
mA	20	20	80	80
Ms	20	20	13	13
Cone angle for narrow scan	0.5°	0.5°	0.5°	0.5°
Cone angle for wide scan	10°	10°	9°	9°

Table 2

Comparison of mean reconstruction values in HU inside the contrast rods of the Catphan®500 phantom without a full-fan bowtie.

ROI	Prior	Wide	Corrected	Percent difference between wide and prior	Percent difference between corrected and prior
1	276	170	260	39%	6%
2	85	10	78	88%	8%
3	-859	-740	-782	14%	9%
4	-64	-102	-49	60%	22%
5	-110	-138	-92	25%	17%
6	-185	-196	-159	6%	14%
7	-857	-731	-774	15%	10%
8	813	583	740	28%	9%
9	42	-48	37	213%	13%
Average				54%	12%

Table 3

Comparison of mean reconstruction values in HU inside the contrast rods of the Catphan[®]500 phantom with a full-fan bowtie.

	Prior	Wide	Corrected	Percent difference between wide and prior	Percent difference between corrected and prior
1	284	190	269	33%	5%
2	86	29	94	66%	10%
3	-870	-754	-765	13%	12%
4	-54	-85	-32	58%	41%
5	-104	-123	-76	18%	27%
6	-182	-189	-147	4%	19%
7	-870	-754	-766	13%	12%
8	833	612	737	27%	12%
9	101	39	124	62%	22%
Average				33%	18%

Table 4

Expansion coefficients for full-fan mode scans.

Coefficient	Catphan		Rando head		Water phantom	
	No bowtie	Bowtie	No bowtie	Bowtie	No bowtie	Bowtie
a0	-0.0075	0.0513	-0.0007	0.0690	0.0007	0.0253
a1	0.8760	0.8548	0.9998	0.8751	0.9016	0.8474
a2	0.2004	0.2939	0.0412	0.1799	0.1878	0.2261
a3	-0.1026	-0.1491	-0.0281	-0.0863	-0.1104	-0.0867
a4	0.0236	0.0313	0.0091	0.0189	0.0280	0.0145
a5	-0.0019	-0.0023	-0.0009	-0.0015	-0.0024	-0.0009

Table 5

Comparison of mean reconstruction values in HU inside the contrast rods of the density plug phantom without a half-fan bowtie.

	Prior	Wide	Corrected	Percent difference between wide and prior	Percent difference between corrected and prior
1	-94	-293	-66	213%	30%
2	-701	-557	-558	21%	20%
3	-15	-335	-126	2079%	722%
4	-111	-347	-138	211%	24%
5	-117	-334	-130	186%	11%
6	147	-229	91	256%	38%
7	-486	-427	-315	12%	35%
8	152	-196	142	229%	7%
9	-687	-471	-588	31%	14%
10	-474	-423	-415	11%	13%
11	13	-178	-8	1427%	161%
12	19	-146	31	887%	69%
13	-58	-175	0	202%	100%
14	14	-134	30	1081%	117%
15	829	189	582	77%	30%
16	823	251	682	69%	17%
Average				437%	88%

Table 6

Comparison of mean reconstruction values in HU inside the contrast rods of the density plug phantom with a half-fan bowtie.

	Prior	Wide	Corrected	Percent difference between wide and prior	Percent difference between corrected and prior
1	-87	-256	-26	196%	70%
2	-71	-280	-59	293%	17%
3	201	-215	102	207%	49%
4	-458	-422	-316	8%	31%
5	200	-167	207	184%	4%
6	-69	-250	58	262%	184%
7	-686	-500	-494	27%	28%
8	21	-219	62	1154%	200%
9	-730	-539	-603	26%	17%
10	-12	-237	-93	1831%	657%
11	-461	-449	-421	2%	9%
12	19	-173	15	1001%	21%
13	20	-127	67	744%	240%
14	-68	174	2	156%	103%
15	745	170	540	77%	28%
16	812	292	662	64%	18%
Average				389%	105%

Table 7

Expansion coefficients in half-fan mode with and without a half-fan bowtie filter.

Coefficient	Density plug		Shoulder		Pelvis	
	No bowtie	Bowtie	No bowtie	Bowtie	No bowtie	Bowtie
a0	0.0086	0.0000	0.0092	0.0185	0.0059	0.5133
a1	0.9501	1.1298	0.8537	0.9378	0.8838	0.8649
a2	0.1239	0.2266	0.2560	0.0944	0.2134	0.1568
a3	-0.0589	0.1911	-0.1466	-0.0083	-0.1198	-0.0270
a4	0.0108	-0.0543	0.0370	-0.0040	0.0283	-0.0030
a5	-0.0003	0.0053	-0.0031	0.0007	-0.0022	0.0002