Correction of streaking artifacts in CT images and its influence on Monte Carlo dose calculations

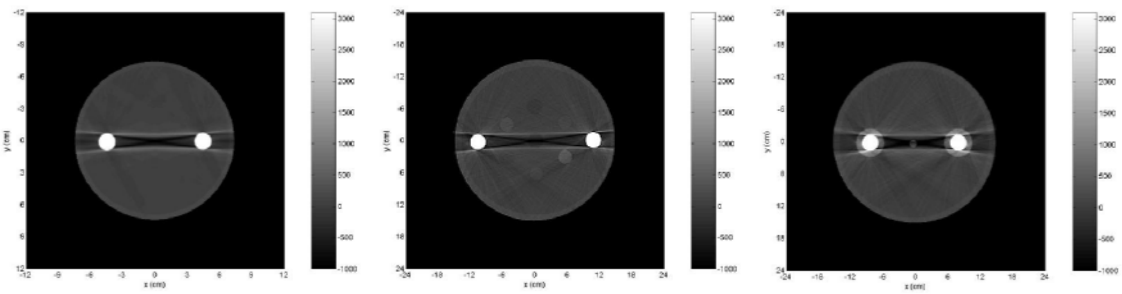
Introduction: Currently, treatment planning is done on the basis of CT (computed tomography) images from which a patient density matrix is determined. In conventional treatment planning the dose is calculated assuming patients are composed of water, at best considering electron densities. In comparison to that, in Monte Carlo Treatment Planning (MCTP), the CT image is segmented into a few materials (e.g. air, bone, tissue), and the dose is calculated taking into account these media. If there is an inaccuracy in the conversion of HU (Hounsfield units) to material type and density in the CT image, there might appear a significant dose miscalculation in both treatment planning methods.

Streaking artifacts are one type of CT artifacts which are caused by very high attenuation of X-rays in some materials, such as metals. For example, between two metallic objects in a patient's body there is practically no information about the patient's geometry which can be detrimental for treatment planning. Such artifacts occur in prostate patients with two hip implants or for head and neck patients with tooth fillings.

The latest approach of CT artifact correction deals with raw data - sinograms - and filling of missing projections corresponding to rays that passed through the highly attenuating media.

In this work, a method for streaking artifact correction in sinogram space has been developed and the method was validated by dose calculation for MV photon beams.

Materials and Methods: Three phantoms were used for the purpose of our study, as depicted in Fig. 1. All phantoms were filled with water and the pelvic phantom contained several materials of different contrasts. In CT images of all three phantoms, there is a significant image distortion due to streaking artifacts caused by the presence of the steel cylinders (Fig. 1).

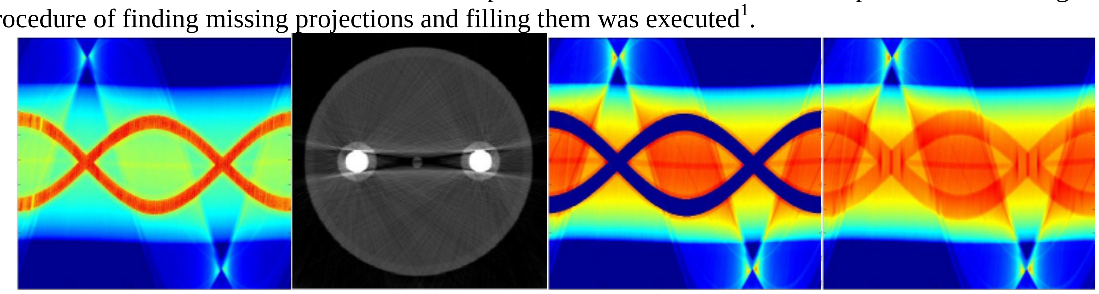


a **b** **c**

Figure 1: (a) Streaking artifacts in a 15 cm diameter head phantom with two metal implants modeling dental fillings, (b) in a 27 cm diameter pelvic contrast phantom (#1) with simple metal implants, and (c) in the same phantom with metal implants surrounded by bone (#2).

The CT images illustrate the need for correction for artifacts. In a case of a prostate patient with two metallic hip implants, the artifacts may cause serious difficulties in delineation of critical organs or the tumor.

Artifact correction: Our test phantoms were scanned with a third generation Picker PQ5000 CT helical scanner, at 120kVp and 400 mAs. Fan beam sinograms were converted into parallel beam sinograms to allow the inverse Radon transforms to be computed in a Matlab routine. In the parallel beam sinogram a procedure of finding missing projections and filling them was executed[1].



a **b** **c** **d**

Figure 2: The steps of artifact correction on the pelvic phantom #2 as described in text: (a) the original sinogram, (b) the original image with artifacts, (c) the original sinogram with the mask, (d) the interpolated sinogram. The CT artifact corrected image reconstructed from the interpolated sinogram is shown in Fig. 3.

First, the image with artifacts (Fig. 2b) was reconstructed from the original sinogram (Fig. 2a), so that highly attenuated objects could be found. A fixed threshold for selecting these objects was used. After that, Radon transform of the metallic objects only was performed. As a result, the sinogram of the metallic objects was used as a mask for the original sinogram and the missing data were identified (Fig. 2c). These data have to be interpolated using information of neighboring rays and/or projections (Fig. 2d).
In this work, the missing data were interpolated within each projection with a cubic spline function. Finally, the image corrected for streaking artifacts can be reconstructed from the modified sinogram. Fig. 3 presents CT artifact corrected images from Fig. 1.

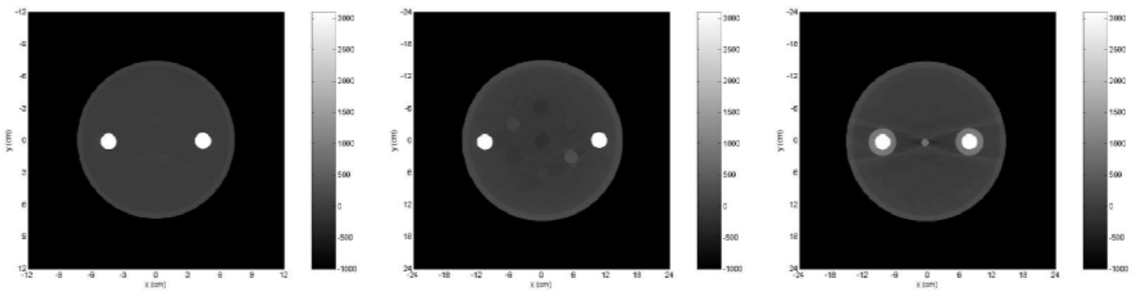


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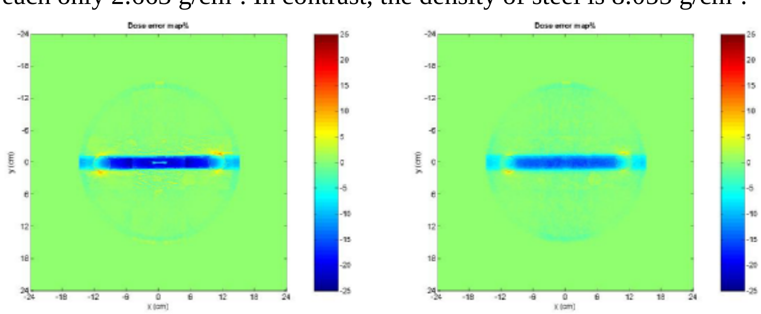
Figure 3: CT artifact corrected images of (a) the head phantom, (b) the pelvic phantom #1, and (c) the pelvic phantom #2.

MC dose calculations: Conventionally, MCTP converts HU into material densities according to a calibration curve and then it segments CT images in a number of materials. In CT images with artifacts, MCTP would fail in assigning media which would result in large errors in dose calculations.
The dose calculations were performed using the EGSnrc/DOSXYZnrc MC code. Two photon beam energies (6 and 18 MV) in two beam arrangements (parallel opposed, four field box) were used.
First, the standard calibration curve for HU-density conversion was used, which segments CT images in four media: air, lung, tissue, and bone. The densities are assigned according to linear interpolations between calibration points (see Table 1).

Material	HU interval	ρ interval	Material	HU interval	ρ interval
Air	[-1000 : -950]	[0.001 : 0.044]	ICRP cortical bone	[125 : 2000]	[1.101 : 2.088]
Lung	[-950 : -700]	[0.044 : 0.302]	**Steel**	**3095**	**8.055**
ICRU tissue	[-700 : 125]	[0.302 : 1.101]			

Table 1: The default DOSXYZnrc calibration with an added point for a metallic object – the **extended calibration**.

It was found that the default calibration gives large errors in between the steel cylinders (Fig. 4) due to two facts: the steel cylinders are assigned to bone and the HU number scale extends only from -1000 to 3095. The latter means that according to the default DOSXYZnrc calibration, the maximum assigned density can reach only 2.663 g/cm^3. In contrast, the density of steel is 8.055 g/cm^3.



a **b**

Figure 4: Dose error maps (D_{exact} – D_{CT})/D_{exact} for the **default calibration** (a) in the original CT image and (b) in the CT artifact corrected image.

The errors in MC dose calculation between the steel cylinders are large: up to 25% in the original image and 18% in the CT artifact corrected image. To take the attenuation in the steel cylinders into account, a metallic material, in addition to the default materials was used in our MC calculations. The default DOSXYZnrc calibration was modified and a new calibration, which we call **extended calibration** (Table 1), was created.

Results: The CT images were segmented into materials according to the extended calibration described above. Material segmentations of the pelvic phantom #1 in the original and in the artifact corrected CT images are presented in Fig. 5.

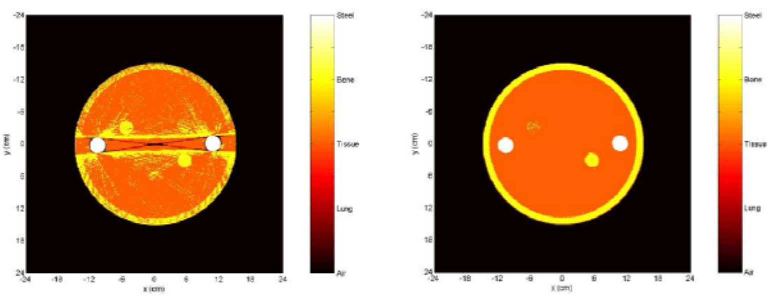


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Figure 5: Material segmentation with the **extended calibration** in pelvic phantom #1 done in (a) original CT image and (b) in the artifact corrected one.

An obvious reduction in image artifacts is apparent in the CT artifact corrected image (Fig. 5a) compared to the original one (Fig 5b). Incorrect medium assignment in the original CT image resulted in voxels between the steel cylinders being assigned as air and lung. The CT image was also very noisy in the central area. On the other hand, in the CT artifact corrected image, the acrylic cylinder with density of 1.19 g/cm^3 was correctly assigned to bone and water was assigned to tissue, as requested by the calibration curve. There is a slight inaccuracy in the assignment of the left upper vial in the CT artifact corrected image. The density of the water solution of $Ca(ClO_4)_2$ is 1.119 g/cm^3 which is very close to the tissue-bone calibration point. Therefore, there are some voxels assigned to bone and some are assigned to tissue. Similar results were obtained for the other two test phantoms.

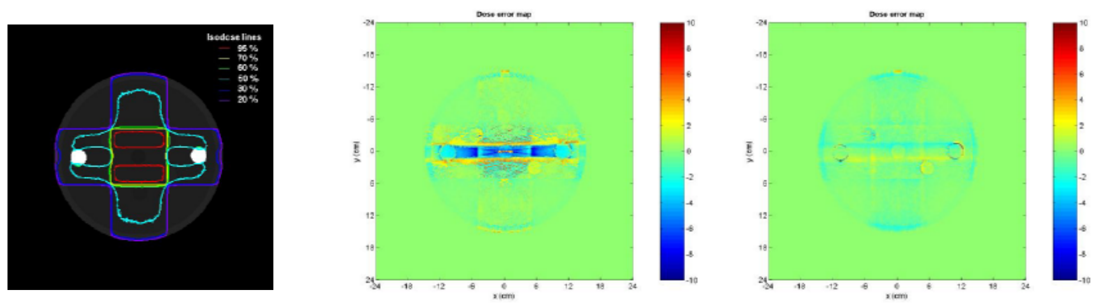

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Figure 6: The isodose distribution computed with the **extended calibration** in (a) the exact geometry of pelvic phantom #1 for four 18MV photon beams. The error in dose calculation (in percent) in (b) the original CT image and (c) the CT artifact corrected image.

The streaking artifacts play a big role in MC dose calculation, as is illustrated in Fig. 6. The significant improvement in dose calculation in the CT artifact corrected image using the extended calibration is evident. The error in the central part of the area of the phantom was decreased from 10% to 2%.

To model a patient pelvic treatment, a target was delineated in the center of two steel cylinders (red line in Fig. 7). This represents the worst case scenario, e.g. prostate cancer treatment for a patient with two bilateral metal hip prostheses. We found that in the original CT image, there are 57% of voxels in the target having dose calculation error larger than 2%, whereas in the CT artifact corrected image, there are only 2% of these voxels. Similar improvement was obtained for dose calculation in the other beam setups.

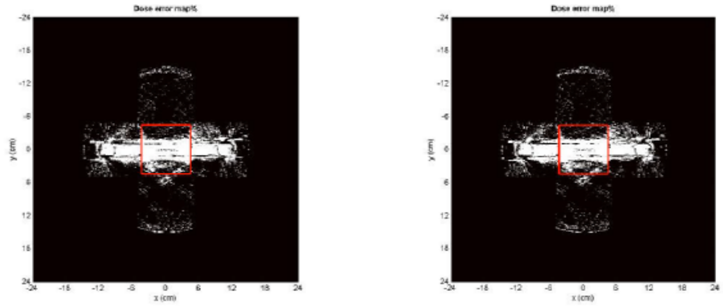

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Figure 7: Map of pixels having dose error larger than 2% in the original CT image (a) and in the CT artifact corrected image (b).

To point out the importance of using the extended calibration, the DVH of the target was plotted for the exact geometry, for original and CT artifact corrected images with default and extended calibration (Fig. 8). The DVH plot shows that changing the default DOSXYZnrc calibration line makes a larger difference than using the artifact correction. In the exact geometry, there is a cold spot in between the prostheses one has to be aware of. This turns into a hot spot when using the default calibration. The importance of adding the extra calibration point is obvious; the DVH curves obtained from MC dose calculations, that use the extended calibration, approach the true DVH. And yet, the DVH curve that was calculated from CT artifact uncorrected image using the extended calibration is significantly different from the exact DVH. A good agreement of DVH curves is reached only when the dose is calculated with the extended calibration in the CT artifact corrected image (yellow line in Fig. 8). The fact that the yellow and the red (the true) DVH curves are so close illustrates the significance of CT artifact correction algorithm.

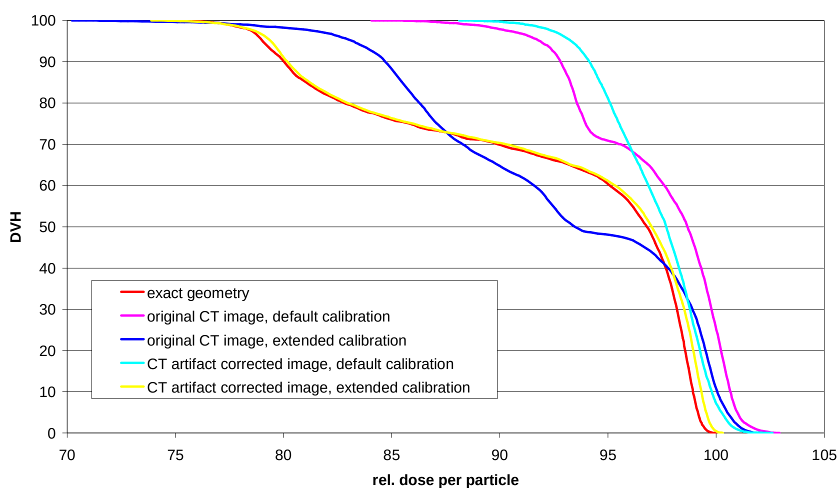

Figure 8: DVH of the target for the exact geometry, and different combinations of corrections. The dose is normalized to the maximum dose per particle delivered in the exact geometry.

Conclusion: A method of correction of streaking artifacts was developed and validated on three test phantoms. MC dose distributions have been calculated for original CT images and CT artifact corrected images with the EGSnrc/DOSXYZnrc code. The calculations were compared to dose calculation in the exact phantom geometry. Image quality is greatly improved by the artifact correction especially in between metallic objects, and makes material segmentation and dose calculation more accurate.

It has been found that in order to calculate dose accurately in between two metallic objects, one has to modify the default DOSXYZnrc calibration curve by adding an extra high density material.

The error in dose calculations improved from 25% for images with CT artifacts when the default CT calibration was used to less than 2% for CT artifact corrected images if the extended calibration was used.

References:

[1] Yazdi M, Gingras L, Beaulieu L, "An adaptive approach to metal artifact reduction in helical computed tomography for radiation therapy treatment planning: experimental and clinical studies." Int J Radiat Oncol Biol Phys **62**(4), 1224-31 (2005).