Interior photon counting computed tomography for quantification of coronary artery calcium: pre-clinical phantom study

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Abstract

Computed tomography (CT) is the reference method for cardiac imaging, but concerns have been raised regarding the radiation dose of CT examinations. Recently, photon counting detectors (PCDs) and interior tomography, in which the radiation beam is limited to the organ-of-interest, have been suggested for patient dose reduction. In this study, we investigated interior PCD-CT (iPCD-CT) for non-enhanced quantification of coronary artery calcium (CAC) using an anthropomorphic torso phantom and ex vivo coronary artery samples. We reconstructed the iPCD-CT measurements with filtered back projection (FBP), iterative total variation (TV) regularization, padded FBP, and adaptively detruncated FBP and adaptively detruncated TV. We compared the organ doses between conventional CT and iPCD-CT geometries, assessed the truncation and cupping artifacts with iPCD-CT, and evaluated the CAC quantification performance of iPCD-CT. With approximately the same effective dose between conventional CT geometry (0.30 mSv) and interior PCD-CT with 10.2 cm field-of-view (0.27 mSv), the organ dose of the heart was increased by 52.3% with interior PCD-CT when compared to CT. Conversely, the organ doses to peripheral and radiosensitive organs, such as the stomach (55.0% reduction), were often reduced with interior PCD-CT. FBP and TV did not sufficiently reduce the truncation artifact, whereas padded FBP and adaptively detruncated FBP and TV yielded satisfactory truncation artifact reduction. Notably, the adaptive detruncation algorithm reduced truncation artifacts effectively when it was combined with reconstruction detrending. With this approach, the CAC quantification accuracy was good, and the coronary artery disease grade reclassification rate was particularly low (5.6%). Thus, our results confirm that CAC quantification can be performed with the interior CT geometry, that the artifacts are effectively reduced with suitable interior reconstruction methods, and that interior tomography provides efficient patient dose reduction.

1. Introduction

Computed tomography (CT) is an important tool in medical diagnostics. Although the harmfulness of effective radiation doses within few millisieverts range encountered in typical CT examinations remains controversial (McCollough et al 2012), the ALARA (As Low As Reasonably Achievable) principle continues to motivate the reduction of effective doses in clinical protocols towards the submillisievert range (McCollough et al 2012). Consequently, the radiation exposure in CT examinations has been reduced over the years through the introduction of automatic exposure control, tube current modulation, tube potential and filtration selection, beam-shaping filters, and Z-axis collimation (McCollough et al 2012). More recently, the commercial introduction of iterative reconstruction (IR) (Beister et al 2012) and deep learning image...
reconstruction algorithms, such as the AiCE (Advanced Intelligent Clear-IQ Engine; Canon Medical Systems, Otawara, Japan) and TrueFidelity™ (GE Healthcare, Waukesha, WI, USA), into diagnostic CT systems has also reduced the radiation dose substantially. Yet, the latest hardware and software innovations, such as photon counting detectors (PCDs) and interior reconstruction, likely continue the trend of reducing radiation doses in future CT systems (McCollough et al 2012).

In interior tomography, the radiation beam is collimated to the volume-of-interest (VOI), which reduces the radiation dose to diagnostically non-relevant peripheral organs and tissues (Natterer 2001). In addition to the dose-reductions in peripheral structures, the highly collimated beam will greatly reduce x-ray scattering from the patient, possibly yielding contrast improvements. However, prominent truncation artifacts arise from the strongly collimated measurement geometry when compared to full field-of-view measurement. Therefore, significant effort has been made in order to build the mathematical basis for interior reconstruction and the reduction of truncation artifacts (Natterer 2001, Noo et al 2004, Wang 2011, Xu et al 2011).

The interior tomography problem is ill-posed and does not have a uniquely determined solution (Natterer 2001). Nonetheless, the introduction of additional a priori information in the reconstruction problem has led to theoretically exact interior reconstructions. For instance, the differentiuated back-projection reconstructs the VOI accurately when a sub-region within the VOI is known a priori (Kudo et al 2008). Conversely, compressed sensing (CS) and IR methods have been successfully applied to solve the interior tomography problem exactly without the need for a known sub-region (Yu and Wang 2009).

With CS-based approaches, the VOI is often assumed piecewise constant, which allows exact reconstruction with the use of some sparsifying transform, e.g., wavelet transform or total variation (TV) minimization (Yu and Wang 2009). This flexibility of CS algorithms is highly desirable as these methods could be directly applied to virtually any imaging scenario without the need for prior information about the object. In addition to CS approaches for interior reconstruction, several sinogram extension techniques have been implemented. Generally, these approaches estimate the missing part of the sinogram by padding the boundary value (Kyriileis et al 2010), by smoothly damping the sinogram values outside the VOI to zero (Ohnesorge et al 2000), or by estimating the convex hull of the patient and extrapolating the sinogram with this shape-estimate (Sourbelle et al 2005, Taguchi et al 2011).

In this study, we investigated the image quality with the interior tomography imaging geometry using an experimental interior photon counting detector CT (iPCD-CT) setup in the context of cardiac imaging. Truncation artifacts with several interior reconstruction techniques were compared, and to mimic a diagnostic imaging task, the coronary artery calcium (CAC) quantification accuracy of our iPCD-CT system was evaluated using ex vivo coronary artery samples that were positioned in an anthropomorphic torso phantom. Finally, we compared organ doses between iPCD-CT and conventional clinical CT to assess the distribution of radiation dose between these imaging geometries.

2. Materials and methods

2.1. Coronary artery samples and imaging phantom

We extracted coronary artery samples (N = 18) with an approximate length of 2 cm from the left anterior descending, left circumflex, and right coronary arteries from 9 cadavers during a routine medical autopsy. Samples were inserted in a water-filled sample holder within a custom epoxy resin cylinder phantom (9 cm diameter) (figure 1(a)), and they were scanned with our experimental PCD-CT setup twice (see section 2.2). In the first scan, hereafter referred to as the PCD-CT measurement (figure 1(b)), the samples were measured only using the epoxy resin cylinder to avoid the interior tomography problem. In the second scan, hereafter referred to as the iPCD-CT measurement, the epoxy resin cylinder was positioned in a torso phantom (008C, CIRS, Inc., Norfolk, VA) (figure 1(a)) and re-measured using the interior imaging geometry (figures 1(c) and 2(a)). The research protocol was approved by the Ethics Committee of Northern Ostrobothnia Hospital District (Permission number: 40/2018).

2.2. Experimental PCD-CT system and calibration

We used a flat-panel photon counting detector (Flite FX15, XCounter AB, Danderyd, Sweden) in our experimental table-top cone-beam CT system (referred to as the PCD-CT setup) (figure 2(a)). The PCD has a pixel pitch of 100 μm, and it is constructed of 24 separate tiles, each with 256 × 128-pixel area (height × width). The active detector panel area is 5.13 cm × 15.47 cm (height × width). As the x-ray source for our setup, we utilized a C-arm (Philips BV29, Philips Healthcare, Netherlands), with inherent filtration equivalent to 3.0 mm of Aluminum at 75 kVp. The source, with a 0.6 mm focal spot, could not provide the routine CAC scoring peak kilovoltage of 120 kVp and was therefore operated at 100 kVp and 3.0 mA. The PCD, x-ray source, and the motorized rotation stage (NR360S/M, Thorlabs, Inc., Newton, New Jersey) were aligned with a laser pointer. The x-ray source was 88.9 cm and 58.8 cm from the detector and center-of-rotation of the object, respectively (figures 1(b)–(c)). The angular velocity of the rotation stage was fixed at 1.5° s⁻¹, while the PCD was operated at 15 frames/s, allowing 10 frames per angular degree. In order to match the exposures and...
effective doses between CT and iPCD-CT, we selected the number of averaged frames that produced the most comparable effective dose between the two geometries. Consequently, every tenth frame of the PCD-CT and iPCD-CT measurement was used for reconstruction. The energy threshold of the PCD was set at 10 keV to remove electronic noise. The main imaging parameters have been summarized in table 1.

PCDs have been reported to produce inhomogeneous responses between different detector tiles (Ahmad et al 2017, Juntunen et al 2020). We corrected these variations with the signal-to-equivalent-thickness calibration (STC) (Vavrik et al 2006, Jakubek 2007) using polymethyl methacrylate (PMMA) as the calibration material. PMMA has similar mass-attenuation and scattering properties to soft tissues and was therefore considered a suitable calibration material. We measured PMMA thicknesses 0 cm, 5.3 cm, 10.6 cm, 15.9 cm, 21.2 cm, 26.5 cm, and 31.8 cm for STC.

2.3. Truncation correction in projection domain
Since the adaptive sinogram detruncation algorithm has provided excellent interior reconstruction quality within the VOI in previous studies, we decided to apply it for truncation correction (Sourbelle et al 2005, Kolditz et al 2011). In summary, the approach consists of two main steps:

1. Patient shape estimation with an ellipse. For each projection angle, we estimated the minimum amount of PMMA in the STC sinogram (figure 3(a)). The angle with maximum thickness corresponded to lateral view, whereas the projections with the minimum thicknesses were from the anterior and posterior views. Once the orientation of the phantom was known, we created an elliptical mask with major and minor axes corresponding to the measured PMMA thicknesses in lateral and posterior projections. The interior VOI was subsequently removed from the
obtained ellipse mask yielding the torso mask (figure 3(c)).

(2) Elliptical sinogram extrapolation. For each projection angle, the truncated sinogram was extrapolated by fitting separate square root functions \( \sqrt{a\xi^2 + b\xi + c} \), where \( \xi \) is the detector pixel location, and \( a, b, \) and \( c \) are fit parameters) to the left and right borders of the truncated sinogram. The extrapolated sinogram was fitted by ensuring consistency of sinogram values and sinogram gradient values between the extrapolated sinogram and the measured sinogram at the sinogram border. The estimated patient shape from step 1) provided a pixel location (\( \xi \)) for the root value of the extrapolation function (\( \sqrt{a\xi^2 + b\xi + c} = 0 \)), and consequently, we had a system of three equations allowing the determination of the three fit parameters \( a, b, \) and \( c \) (figure 3(d)).

As an alternative approach, we utilized the sinogram padding technique to assess the effect of missing shape information. With this method, the boundary pixel value of the truncated sinogram is repeated to obtain the extended sinogram (figure 3(b)) (Kyrieleis et al 2010).

2.4. Image reconstruction
In order to generate a computational CT projector for calculating the reconstructions, we utilized the ASTRA (v. 1.8, iMinds-Vision Lab, University of Antwerp, Belgium) (van Aarle et al 2015, 2016) and SPOT (v.1.2) toolboxes in MATLAB (v. 9.4, The MathWorks Inc., Natick, MA, 2018). We reconstructed the projection data with FBP, padded FBP, and FBP with adaptive detruncation. Furthermore, since total variation can provide exact reconstructions if the VOI is piecewise constant (Yu and Wang 2009), we decided also to use TV for reconstruction. The slices were reconstructed using the fan-beam geometry in 2D.

As the IR algorithm, we used penalized least squares with TV regularization. The loss function was formulated as

\[
\arg\min_u \frac{\|Au - f\|^2}{2} + \lambda \int_\Omega \sqrt{\|\Delta u\|^2 + \varepsilon} \, dx, \tag{1}
\]

where \( u \) is the reconstructed image, \( f \) is the sinogram, \( A \) is the forward projection matrix, \( \varepsilon \) (\( =10^{-8} \)) is the smoothing parameter, \( \Omega \) is the reconstruction domain, \( x \) is the reconstruction coordinate, and \( \lambda \) is the regularization parameter. The gradient \( \nabla u \) was calculated using the forward-difference method, and the TV-regularized reconstruction was obtained by
Table 1. Imaging parameters for PCD-CT, iPCD-CT, and clinical CT. The clinical CT values have been taken from the protocol at the Oulu University Hospital. CT protocol parameters were selected for the phantom using a demonstration heartbeat setting.

| Parameter     | kVp | Filtration  | Scan time (s) | Exposure (mAs) | Slice thickness (mm) | Field-of-view (cm) | Voxel size | Collimation (mm) | Scanning mode | Pitch |
|---------------|-----|-------------|---------------|----------------|----------------------|---------------------|------------|------------------|---------------|-------|
| PCD-CT/iPCD-CT| 100 | 3.0 mm Al   | 240           | 720            | 3.0                  | 10.2                | 0.32 mm × 0.32 mm × 1.5 mm | 38.4 ± 1°     | Axial | — |
| Clinical CT   | 120 | 0.3 mm Ti + 1.0 mm C + 0.5 mm Al | 0.285 78 | 3.0 | 30 | 0.32 mm × 0.32 mm × 1.5 mm | 38.4 | Helical | 3.4 |

* The collimation for PCD-CT is reported as mean ± error since it was estimated from the measured projection data.
minimizing (1) using the gradient descent method. 500 gradient descent iterations with the Barzilai-Borwein step size update was used to yield the reconstruction (Barzilai and Borwein 1988).

To maintain consistency between iPCD-CT and PCD-CT data, we used the same FBP reconstruction kernel (Hann filter and moving average filter with 4.2-pixel window) for both reconstructions. The TV regularization parameter of 25 was selected to produce comparable noise magnitude with FBP. The cardiac rod phantom was reconstructed in the PCD-CT geometry with FBP, and this reconstruction was used as the ground-truth reconstruction. The reconstructions were converted to Hounsfield units (HUs), by measuring the attenuation value of the water-filled sample holder within the rod phantom (figure 4).

In our preliminary tests, we observed an artifact arising from the strong beam hardening originating from the spine in the iPCD-CT reconstructions, which was not entirely corrected by the sinogram extension techniques. Therefore, as a post-processing scheme, we applied detrending on the reconstructed VOI by, first, blurring the image with a Gaussian filter with a standard deviation of 4, and second, fitting a fourth-order polynomial surface to the filtered image. This polynomial surface was then subtracted from the reconstructions to obtain the detrended slice.

2.5. Image quality evaluation
To assess the extent of truncation artifacts with different iPCD-CT reconstruction methods, we compared their line profiles within the VOI by measuring the HUs of vertical and horizontal lines with a length of 200 pixels. To further assess the extent of truncation, we compared the mean HU values of peripheral regions-of-interest (ROIs) and reported the standard deviation (SD) of these mean values for each reconstruction method. The ROIs were obtained by sampling four squares with a 32 × 32-pixel area (figure 4).

2.6. Quantification of coronary artery calcium
Agatston score is routinely evaluated in the diagnosis of coronary artery disease (CAD) (Agatston et al 1990). It is calculated from a non-contrast enhanced CT scan, and it helps to classify coronary artery disease into no CAD, minimal, mild, moderate, or severe CAD, depending on the volume and the density of the calcification (Knipe and Ayush 2016). Since Agatston scoring is generally performed with 120 kVp, we had to change the segmentation threshold of 130 HU to 147 HU in order to conform with the 100 kVp measurement (Thomas et al 2006, Nakazato et al 2009) (table 2). This Agatston score-analogous CAC score was calculated for each slice, and the sum of the slicewise scores (and volumes) was evaluated for each
artery sample. We used the regression slope, Pearson correlation \((r\text{-} and \; p\text{-}values)\), and Bland-Altman (BA) plots to compare the measured CAC scores and calcium volumes between PCD-CT and iPCD-CT. The non-parametric Wilcoxon signed-rank test (significance level \(p < 0.05\)) was used to evaluate whether there were statistically significant differences in the calcification volumes and CAC scores between PCD-CT and iPCD-CT geometries. The statistical tests were evaluated with MATLAB. The reclassification rate of the CAD grade with the iPCD-CT geometry was also reported using CAD grades from the PCD-CT reconstruction as ground truth values.

### 2.7. Measurements of organ dose for clinical CT and iPCD-CT

We evaluated the organ doses using the approach presented in (Zhang et al 2013). Briefly, we positioned 74 radiophotoluminescence (RPL) dosimeters with a tin filter (GDM-352M, Asahi Techno Glass Corporation, Chiba, Japan) in the dosimeter holes of an anthropomorphic adult male phantom (ATOM 701-C, CIRS Inc., Norfolk, VA), and measured the dosimeter reading for both clinical CT and iPCD-CT. We used a Doce Ace FGD-1000 (Asahi Techno Glass Corporation, Chiba, Japan) for the dosimeter readout. The dosimeter readings for heart, lungs, esophagus, spinal cord, spleen, stomach, thymus, liver, thyroid, and kidneys were measured. To determine the effective dose, we used the tissue weighting factors from (ICRP 2007) for these organs and set the radiation doses of other organs to zero. For iPCD-CT reconstruction, we utilized only every tenth frame, and consequently, the measured organ doses for iPCD-CT were divided by ten to yield the organ doses for the iPCD-CT protocol. As our experimental PCD-CT setup is not representative of the clinical CT scan, we decided to measure the reference CT organ doses using the calcium scoring protocol of a clinical CT scanner (figure 2(b)) (Somatom Definition Flash, Siemens Healthcare, Erlangen, Germany). The main imaging parameters for this clinical CAC scoring protocol have been summarized in (table 1). However, to make the x-ray sources between clinical CT and iPCD-CT more comparable, we reduced the peak kilovoltage of the clinical CT tube from the conventional calcium scoring value of 120 kVp to 100 kVp as it was also used for iPCD-CT.

The dosimeters for organ dose measurements are usually calibrated to the dosimeter reading in the air \((D_{air})\). Therefore, to obtain the energy deposited to the tissue (organ dose), we used the correction (Hendee and Ritenour 2002)

\[
D_{tissue} = D_{air} \times \left( \frac{\mu_{tissue}(E_{eff})}{\mu_{air}(E_{eff})} \right)_{tissue}\left( \frac{\mu_{air}(E_{eff})}{\mu_{air}(E_{eff})} \right)_{air}
\]

where \(E_{eff}\) is the effective photon energy, \(\rho\) is the density of the material, and \(\mu_{tissue}\) is the mass energy-absorption coefficient (Hubbell and Seltzer 2004). We simulated the x-ray spectra of the clinical CT and iPCD-CT systems using the Spektripaja-software (Tapiokaara and Tapiokaara 2007) and obtained effective energy of 50.8 keV for both CT and iPCD-CT. Since we do not compare the image quality between clinical CT and iPCD-CT, we matched their effective doses and focused on analyzing the redistribution of organ doses with iPCD-CT. We also reported the effective doses to represent the stochastic health risk of the CT and iPCD-CT protocols.

### 3. Results

#### 3.1. Image quality with inferior tomography

Cupping artifacts were observed in both horizontal and vertical line profiles for FBP, but the strong attenuation of the spine induced visible cupping only in the horizontal line profile of the TV reconstructions (figure 5). Padded FBP, on the other hand, over-corrected the truncation artifact causing a slight inverse cupping effect in the line profiles (figure 5). The detuncated FBP and TV reconstructions showed the smallest deviation from the ground truth reconstruction of the rod and the smallest cupping in the vertical direction. Nonetheless, the strong attenuation of the spine was not accurately corrected with the detruncation algorithm, resulting in horizontal line profiles whose HU values gradually increased when moving closer to the spine (figure 5). Furthermore, the beam hardening of the ribs induced a dark band within the VOIs of each reconstruction (figures 5 and 6, white rectangle). The detrending resolved the truncation caused by the spine by removing the cupping patterns in both directions for padded FBP and detruncated reconstructions (figure 6). The combination of adaptive sinogram detruncation and reconstruction detrending provided excellent cupping removal and the smallest standard deviation in the measured mean ROI values between the different ROI locations (table 3). Thus, we limited the CAC scoring analysis for the detruncated FBP reconstructions with and without reconstruction detrending.

### Table 2. Agatston score for coronary artery disease grading.

| Maximum HU | Density factor | Agatston score | CAD grade |
|------------|----------------|----------------|-----------|
| <147       | 0              | 0              | No evidence of CAD |
| 147–199    | 1              | 1–10           | Minimal   |
| 200–299    | 2              | 11–100         | Mild      |
| 300–399    | 3              | 101–400        | Moderate  |
| ≥400       | 4              | >400           | Severe    |

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3.2. Calcium scoring accuracy

The calcification volumes and scores ranged between 0 mm$^3$–634 mm$^3$ and 0–1691, respectively, and the collected artery samples featured every plausible CAD grade (no CAD: 2 samples, minimal CAD: 2 samples, mild CAD: 4 samples, moderate CAD: 3 samples, severe CAD: 7 samples). We found strong correlations for both calcium volumes and CAC scores between iPCD-CT and PCD-CT geometries (table 4). Nonetheless, the regression slope values and BA-plots for detruncated FBP and detruncated and subsequently detrended FBP indicated underestimation of both calcium volumes and scores with the iPCD-CT geometry (table 4 and figure 7). Because of this underestimation, we observed a clear increase in the HUs of FBP and TV (pixel positions 100–200 in the horizontal line profile), while this phenomenon is reduced with padded and detruncated reconstructions. The white square indicates the dark band caused by the beam hardening in the ribs located outside the volume-of-interest.

3.3. Organ doses

The effective dose of the iPCD-CT protocol (0.27 mSv) was approximately the same as that of clinical CT (0.30 mSv) (table 5). With a similar effective dose, the organ dose of the heart was increased by 52.3% with iPCD-CT when compared to CT. On the other hand, the radiation strain to more peripheral radiosensitive organs was often reduced (table 5). For the stomach, as an example, the organ dose was reduced by 55.0%.

4. Discussion

In this study, we investigated the applicability of interior tomography imaging geometry in the context of cardiac imaging with an experimental photon counting CT setup. Truncation artifacts and the quantification of coronary artery calcium between PCD-CT and interior PCD-CT were compared. The preservation of HU values in interior reconstructions was observed to be challenging: the inclusion of a priori information on the shape of the torso in the adaptive detruncation algorithm improved the overall image quality but did not completely remove the artifacts caused by the strong attenuation of the spine and ribs. Therefore, strongly attenuating objects outside the reconstruction VOI may cause artifacts that...
are not fully corrected with these approaches. Consequently, detrending had to be applied to the reconstructions to remove the truncation caused by the spine. Finally, the radiation was more effectively focused on the heart with the interior PCD-CT geometry compared to conventional CT. Accordingly, the organ doses of some diagnostically non-relevant organs in cardiac CT were effectively reduced with interior tomography.

The Bland-Altman plots illustrated an underestimation of calcium volume and calcium score with the interior PCD-CT geometry, which was likely related to the increased beam-hardening with the torso phantom. Since the torso attenuates substantially more than the rod phantom, the beam hardens, and the HU-values of the calcifications decrease. This reduced calcification brightness in iPCD-CT reconstructions was evident when visually comparing the PCD-CT and iPCD-CT reconstructions (figures 5 and 6). As a fixed segmentation threshold is used in calcium scoring, the reduced calcium brightness directly affected the segmentation of the calcium border, and consequently, the CAC volumes and scores were lower with iPCD-CT. Despite this phenomenon, the reclassification rate of CAD grades with iPCD-CT (5.6%) was within an acceptable range, since, for example, a change in the scan starting position induces CAD grade reclassification in 10%–11% of cases (Rutten et al 2008, Devries et al 1995).

Interior PCD-CT reduced the organ doses of peripheral organs effectively. This finding is consistent with a previous Monte Carlo simulation study, in which interior CT geometry increased the absorbed dose within the VOI while simultaneously decreasing

Table 3. Standard deviation of the measured mean region-of-interest values.

|                | FBP_iPCD-CT | FBP_iPCD-CT | TV_iPCD-CT | Padded FBP_iPCD-CT | Detruncated FBP_iPCD-CT | Detruncated TV_iPCD-CT |
|----------------|-------------|-------------|------------|--------------------|-------------------------|-----------------------|
| No detrending  | 1.1 HU      | 78.9 HU     | 115.1 HU   | 14.3 HU            | 23.0 HU                 | 23.0 HU               |
| With detrending| —           | 19.5 HU     | 25.5 HU    | 19.8 HU            | 12.2 HU                 | 12.5 HU               |

* Photon counting detector computed tomography.

* Interior photon counting detector computed tomography.

Figure 6. Detrended reconstructions of the interior volume-of-interest and the reference PCD-CT reconstruction of the rod with windowing [−500, 750] HU and their respective line profiles. The cupping originating from the sinogram truncation and the strong attenuation of the spine is reduced through detrending.
it in the lateral anatomy outside the VOI (Wang et al. 2019). Since there are several radiosensitive organs in the proximity of the heart, the use of interior tomo-
graphy in cardiac imaging is well reasoned in terms of radiation protection. Because of patient scatter and as the peripheral organs are directly irradiated in some projection views, e.g., in lateral view for lungs, the dose to these organs cannot, however, be eliminated entirely.

In terms of image quality, TV reduced the truncation artifact in the iPCD-CT geometry when compared to FBP. Nevertheless, the truncation caused by strongly attenuating structures outside the interior VOI, i.e., spine, was evident in the TV reconstruction.

Padded FBP reduced this truncation artifact substantially, but it conversely induced small inverse cupping artifact in the reconstructions, which has been reported in previous research (Kyrieleis et al. 2010, Paleo and Mirone 2017). The magnitude of this cupping artifact is dependent on the length of the extended sinogram (Kyrieleis et al. 2010), and a priori information on the object shape would likely reduce the cupping. The adaptive detruncation algorithm assesses this limitation by including a patient shape estimate and extending the sinogram to this estimate. Accordingly, the adaptive detruncation algorithm reduced the truncation and cupping artifacts most effectively and produced visually comparable

| Parameter | Volume | Agatston score |
|-----------|--------|----------------|
| Reconstruction method | Detruncated FBP | Detruncated and detrended FBP | Detruncated FBP | Detruncated and detrended FBP |
| $r$ (slope) | 0.99 (0.87) | 0.99 (0.87) | 0.99 (0.87) | 0.99 (0.88) |
| Wilcoxon $p$ | 0.0019 | 0.0097 | 0.0013 | 0.0072 |

* $p < 0.0001$ significance level for Pearson correlation.

* Statistical difference was tested with non-parametric Wilcoxon signed rank test. Two-tailed $p$-values are listed.

Figure 7. Bland-Altman plots comparing the coronary artery calcium (CAC) volumes and scores obtained with PCD-CT geometry and those with iPCD-CT. (a) Volume quantification accuracy with detruncated FBP, (b) CAC score quantification accuracy with detruncated FBP, (c) Volume quantification accuracy with detruncated and detrended FBP, (d) CAC score quantification accuracy with detruncated and detrended FBP. The shaded areas indicate the confidence interval limits for mean and agreement limits.

Table 4. Volume and Agatston score correlations between ground-truth PCD-CT reconstructions and iPCD-CT geometry reconstructions.
truncation correction to that of the original study (Sourbelle et al 2005).

The HU values between different interior reconstructions were similar at the center of the rod phantom. This results from the HU calibration process in which the insert holder, positioned at the center of the phantom, is used. As the water calibration insert was positioned at the center of the phantom, the HU calibration was optimized for the central part of the rod. Therefore, the HU values were comparable in this region for each reconstruction method. The HU differences in the peripheral parts of the rod, on the other hand, originated from the truncation artifact. This is a critical point to emphasize and consider: since the samples were positioned in the center of the reconstructed region, with optimal HU values, the observed good calcium quantification accuracy is relatively unsurprising. On the other hand, calculations in the peripheral structures of the VOI would be accurately quantified only if the truncation artifact is effectively suppressed. In light of these results, calcium scoring of peripheral calcifications is not feasible with FBP or TV since they exhibited strong truncation. In contrast, the sinogram extension techniques (sinogram padding and adaptive detruncation) combined with reconstruction detrending provided uniform reconstructions that are more suitable for CAC quantification in the peripheral regions of the VOI.

Overall, interior PCD-CT with adaptive sinogram detruncation and reconstruction detrending reached satisfactory image quality. The calcification structure was visually well preserved, and the rod phantom was accurately reconstructed with negligible cupping. Because of the reduced photon flux at the detector when compared to measuring the cardiac rod, the interior reconstructions were noisier compared to PCD-CT reconstructions of the rod. Additionally, the TV reconstructions portrayed the characteristic ‘paint-brushed’ appearance of iterative reconstruction methods (Geyer et al 2015).

Some limitations of this study have to be assessed. Typically 120 kVp CT protocol is used for calcium scoring, but our x-ray source was limited to 100 kVp, and consequently, we had to apply a calcium scoring method that is optimized for 100 kVp tube kilovoltage similarly as presented in (Thomas et al 2006, Nakazato et al 2009). However, excellent CAC scoring accuracy with reduced patient dose has been demonstrated by combining 100 kVp tube voltage with a tin filter and by using a dedicated calcium scoring reconstruction kernel (Tesche 2017). Furthermore, because of the limited active area of our detector, we could not measure the torso phantom in the PCD-CT geometry with the full torso in the field-of-view. Therefore, we measured the cylindrical rod phantom and used it as our PCD-CT reference. Although this measurement provided a high-quality reference, the attenuated photon spectra between PCD-CT and iPCD-CT scans deviated substantially, which influenced the calcium contrast and most likely caused the observed underestimation of CAC volumes and scores with iPCD-CT.

Concerning radiation dose determination, the radiation dose levels were selected to produce a comparable effective dose between CT and iPCD-CT geometries. Alternatively, the protocols could have been optimized to yield a similar organ dose for the heart. Furthermore, the helical scan mode in the clinical CT protocol caused a small surplus radiation dose outside the FOV due to the requirement for overscanning at the start and the end of the helical CT scan. Finally, the number of coronary artery samples (N = 18) was limited, yet the observed strong correlations between PCD-CT and iPCD-CT CAC volumes and scores were clearly illustrated even with this small number of samples.

The translation of these results into future clinical photon counting CT scanners needs to be assessed. In particular, the used PCD has severely limited count rate capabilities and would likely have challenges enduring the flux of a clinical scanner (Taguchi and Iwanczyk 2013). Consequently, as the x-ray tube output might have to be limited with PCDs, cardiac motion and breathing of the patient may introduce further challenges in a clinical setting with prolonged scan times. To improve the detector performance with respect to x-ray flux, dead time free operation mode, in which one counter records signal and the other is read out, has been developed for the Medipix3 PCD (Ballabriga et al 2013). Alternatively, a dynamic bowtie filter could restrain the maximum flux at the detector (Taguchi and Iwanczyk 2013, Huck et al 2019, Shunhavanich et al 2019).

In future studies, the tissue quantification accuracy in peripheral structures of the interior VOI should be investigated in more detail as these regions are more susceptible to truncation artifact and changes in attenuation quantification (HU values). Moreover, the

| Organ    | CT dose (mGy) | iPCD-CT dose (mGy) | Percentage change (%) |
|----------|---------------|--------------------|-----------------------|
| Heart    | 1.53          | 2.33               | 52.3                  |
| Lungs    | 1.11          | 1.02               | −8.1*                 |
| Esophagus| 0.62          | 0.64               | 3.2                   |
| Spinal cord | 0.60      | 0.42               | −30.0                 |
| Spleen   | 0.43          | 0.17               | −60.5                 |
| Stomach  | 0.40          | 0.18               | −55.0                 |
| Thymus   | 0.39          | 0.49               | 25.6                  |
| Liver    | 0.37          | 0.21               | −43.2                 |
| Thyroid  | 0.11          | 0.10               | −9.1                  |
| Kidneys  | 0.12          | 0.05               | −58.3                 |
| Effective dose (mSv) | 0.30        | 0.27               | −10.0                 |

* Negative value in percentage change indicates dose reduction with iPCD-CT.
image quality between CT and interior PCD-CT should be analyzed along with the assessment of organ doses to determine whether interior tomography can reduce the effective dose without compromising the image quality within the field-of-view. Also, these results should be translated to pre-clinical PCD-CT to assess the impact of cardiac motion and high photon flux on reconstruction quality.

5. Conclusion

In conclusion, excellent truncation correction and satisfactory coronary artery calcium volume and score quantification are possible with interior tomography by extending the truncated sinogram with a priori knowledge on the shape of the torso and by subsequently applying the conventional FBP reconstruction algorithm and reconstruction detrending. Furthermore, the measured dose to peripheral organs can be greatly reduced with the interior geometry.

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