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Detection of increased vasa vasorum in artery walls: Improving CT number accuracy using image deconvolution

Kishore Rajendran^a, Shuai Leng^a, Steven M. Jorgensen^b, Dilbar Abdurakhimova^a, Erik L. Ritman^b, and Cynthia H. McCollough^{*,a}

^aDept. of Radiology, Mayo Clinic, 200 First Street SW, Rochester, MN, USA 55905

^bPhysiology and Biomedical Engineering, Mayo Clinic, 200 First Street SW, Rochester, MN, USA 55905

Abstract

Changes in arterial wall perfusion are an indicator of early atherosclerosis. This is characterized by an increased spatial density of vasa vasorum (VV), the micro-vessels that supply oxygen and nutrients to the arterial wall. Detection of increased VV during contrast-enhanced computed tomography (CT) imaging is limited due to contamination from blooming effect from the contrast-enhanced lumen. We report the application of an image deconvolution technique using a measured system point-spread function, on CT data obtained from a photon-counting CT system to reduce blooming and to improve the CT number accuracy of arterial wall, which enhances detection of increased VV. A phantom study was performed to assess the accuracy of the deconvolution technique. A porcine model was created with enhanced VV in one carotid artery; the other carotid artery served as a control. CT images at an energy range of 25–120 keV were reconstructed. CT numbers were measured for multiple locations in the carotid walls and for multiple time points, pre and post contrast injection. The mean CT number in the carotid wall was compared between the left (increased VV) and right (control) carotid arteries. Prior to deconvolution, results showed similar mean CT numbers in the left and right carotid wall due to the contamination from blooming effect, limiting the detection of increased VV in the left carotid artery. After deconvolution, the mean CT number difference between the left and right carotid arteries was substantially increased at all the time points, enabling detection of the increased VV in the artery wall.

Keywords

atherosclerosis; vasa vasorum; x-ray computed tomography; deconvolution; contrast media

INTRODUCTION

An increase in the spatial density of vasa vasorum, the microscopic vessels in artery walls, is known to precede the development of plaques^{1, 2}. Measuring this change may enable early diagnosis of plaques prior to stenosis. The use of photon-counting CT (PCCT) to measure

* McCollough.Cynthia@mayo.edu; Phone: +1 (507) 538-7723, Fax: +1 (507) 266-3661.

the vasa vasorum density can provide a non-invasive diagnostic method. Contrast-enhanced CT using photon-counting detectors has the added advantage of improved contrast-to-noise ratio³ (CNR) as compared to conventional energy-integrating detectors, and has shown promising preclinical results in vascular imaging^{2, 4}. During contrast-enhanced CT of arteries, a small fraction of the contrast media from the artery lumen enters the vasa vasorum in the wall. The CT number of the wall is proportional to the spatial density of the vasa vasorum. However, one of the key challenges to accurate quantification of the wall enhancement is the blooming effect from the contrast-enhanced lumen that causes overestimation of the wall CT number. Blooming is caused by the limited spatial resolution of the imaging system. Given the submillimeter thickness of the arterial wall, it is desirable to minimize the blooming effect in order to accurately measure the wall CT number and quantify the spatial density of vasa vasorum (VV). If the system point spread function (PSF) is known, then the blooming effect can be modeled as a convolution of the PSF and the true signal. This approach is widely reported to reduce calcium blooming from vascular calcifications, and improve lumen morphology assessment^{5, 6}.

The purpose of this study was to improve CT number accuracy of the artery wall using image deconvolution applied to PCCT images of carotid phantom and animal model from a whole-body research PCCT system. This was accomplished by quantitatively assessing the improvements in CT number accuracy in a 3D printed carotid phantom scan and a scan involving a porcine model of increased vasa vasorum density.

METHODS AND MATERIALS

1.1 Photon-counting CT system

We used a whole-body research photon-counting CT scanner⁷⁻⁹ (Siemens Healthcare, Forchheim, Germany) in our study. The scanner consists of two subsystems, one with a conventional energy integrating detector (EID), and the second subsystem with a photon-counting detector (PCD). The in-plane field of view is 500 mm for the EID subsystem and 275 mm for the PCD subsystem. Two acquisition modes for the PCD are available: Macro mode (2 energy thresholds), and chess mode (4 energy thresholds), each with 0.5mm pixel size at the scan iso-center. A detailed description of these operational modes is published elsewhere^{3, 7}. All scans in this study were performed using the macro mode. While scanning objects that exceeded the PCD FOV, an additional low-dose, data-completion scan (DCS) using the EID subsystem was employed to eliminate truncation artifacts¹⁰.

1.2 Image deconvolution

If the PSF of the imaging system is known, then the non-blind image deconvolution is formulated as a minimization optimization problem¹¹ given by

$$\min_g \sum_{i=1}^N \left(\frac{\lambda}{2} (g \otimes h - f)_i^2 + \sum_{j=1}^J |(g \otimes d_j)_i|^\alpha \right) \quad (1)$$

Here $\mathbf{g}$ is the true signal, convolved with the system PSF $\mathbf{h}$, $\mathbf{f}$ is the measurement/observed signal, λ is a regularization weight, $\mathbf{i}$ is the pixel index, $|\cdot|^\alpha$ is the penalty function, and $\mathbf{d}$ is a first-order derivative filter set ($\mathbf{d}_1 = [1, -1]$ and $\mathbf{d}_2 = [-1, 1]$). The derivative filter(s) produce sparse output rendering the problem non-convex. To solve this non-convex optimization problem and improve computational speed, an alternating minimization approach using half-quadratic splitting¹¹ was used. The problem can be reformulated as

$$\min_{\mathbf{g}, \mathbf{w}} \sum_i \left(\frac{\lambda}{2} (\mathbf{g} \otimes \mathbf{h} - \mathbf{f})_i^2 + \frac{\beta}{2} \left(\|D_i^1 \mathbf{g} - \mathbf{w}_i^1\|_2^2 + \|D_i^2 \mathbf{g} - \mathbf{w}_i^2\|_2^2 + |\mathbf{w}_i^1|^\alpha + |\mathbf{w}_i^2|^\alpha \right) \right) \quad (2)$$

where $\mathbf{w}_i^1$ and $\mathbf{w}_i^2$ are auxiliary variables, β is an optimization weight, and $D_i^j \mathbf{g} = (\mathbf{g} \otimes \mathbf{d}_j)_i$. For a fixed β , Eq. 2 can be iteratively solved by alternating and solving for $\mathbf{w}$ given $\mathbf{g}$, and $\mathbf{g}$ given $\mathbf{w}$. Methods to solve this alternating sub-problem optimization can be found elsewhere^{11, 12}.

The convolution kernel $\mathbf{h}$ is measured using a thin wire phantom (0.127 mm steel wire). The wire was aligned the scanner iso-center, parallel to the patient table and imaged. PCCT thresholds were set at 30 and 65 keV in two-counter macro mode, at 140 kV. Axial images from the 30–140 keV acquisition were reconstructed using a medium smooth D30 kernel and used in image deconvolution. PSF images were reconstructed with $0.13 \times 0.13 \times 0.5$ mm voxels. DCS was not required as the PSF phantom was smaller than the in-plane FOV for PCD.

1.3 Phantom experiment

A carotid artery model was obtained using patient images from a contrast-enhanced CT angiogram. This model was 3D-printed using a polymer based material. The lumen of the artery was approximately 5 mm in diameter and the wall was 2 mm thick. The lumen was injected with iodine contrast (Omnipaque 350, GE Healthcare, Inc) at a concentration of 30mg/mL and the sample was placed in a 20 cm water phantom to simulate the attenuation of neck region. The CT scan was performed at 120 kV, 340 mAs (effective) and PCCT energy thresholds were set to 25 and 52 keV in macro mode. Images were reconstructed with a medium smooth, quantitative kernel (D30). The full spectrum image (25–120 keV) was used for image deconvolution. CT numbers in the wall were measured and compared before and after image deconvolution. Since the total phantom diameter was smaller than the PCD's in-plane FOV, no DCS scan was required. Images were reconstructed with $0.13 \times 0.13 \times 0.5$ mm voxels.

CT number was assessed in the phantom wall using annular ROI placed outside the lumen boundary in the wall region. The lumen boundary was approximated using the half-maximum CT number value from the lumen. A one-pixel thick annular ROI was placed at a distance of one pixel outside the lumen boundary, and subsequently traversed radially outward in one pixel increments. The mean CT number of the ROIs placed at different radial distances from the lumen boundary was measured and compared before and after image deconvolution.

1.4 Animal experiment

A porcine model of enhanced VV density originally developed in rabbits was adopted by our group for this study^{2, 13}. The VV enhancement was performed on the left carotid artery (termed as increased VV artery) while the right carotid artery with normal VV density served as the control. The animal was anesthetized (Telazol/Ketamine/Xylazine) and the carotids were exposed via a midline incision. Aliquot (0.1mL) of autologous blood was injected in the left arterial wall at six sequential locations below the carotid bifurcation. The right artery was exposed but not injected. The animal was allowed to recover. For the PCCT scans, the animal was re-anesthetized and a low dose DCS scan (120 kV, 150 mAs) was performed covering the region of interests. Four locations between the bifurcation and a caudal location were selected for PCCT scans. Iodine contrast agent (Omnipaque 350, GE Healthcare, Inc) was injected in the form of a bolus (1mL/kg bodyweight) into the femoral vein at 5 mL/s followed by a 30 mL saline chaser while scanning each of the four locations. Multiple locations along the long axis of the neck were scanned starting at the time of bolus injection every 3s over 60s resulting in a total of 20 time points per location. PCCT scans were performed at 120 kV, 550 mAs and the energy thresholds were set to 25 and 52 keV in macro mode. Images were reconstructed offline using weighted filtered back projection (Siemens Healthcare, Forchheim, Germany) with a medium smooth quantitative kernel (D30). The full spectrum image (25–120 keV) was used for image deconvolution and CT number measurements of artery walls.

For CT number assessment, smaller patches (151×151) corresponding to the left and right carotids were extracted from the images before and after deconvolution. The half-maximum CT number value from the lumen was used as the threshold to generate a binary mask representing the lumen region. The boundary of the lumen was contour-traced, and used as a reference point for ROI generation and placement. Typically, The carotid artery wall thickness is about 30% of the lumen radius^{2, 14} (typically, the lumen is about 3 mm in radius and the upper boundary for wall thickness is 1 mm). We reconstructed images using a small voxel size ($0.13 \times 0.13 \times 0.5$ mm) to ensure that we had about 6 to 9 voxels accommodating the artery wall. In the single slice images, we used a three pixel-wide ROI placed two pixels away from the identified lumen boundary to avoid partial volume averaging from the lumen-wall interface. The segmentation, lumen identification and ROI placements were performed identically but independently for the control and increased VV arteries, before and after image deconvolution.

RESULTS

The results of image deconvolution applied to a reconstructed PCCT phantom image are shown in Figure 1. A noticeable reduction in blooming effect was demonstrated, and a clear delineation between the wall and the lumen was achieved after deconvolution.

Figure 2 shows the plot corresponding to the line profile shown in Figure 1 and mean CT number from ROIs placed at radial distances from the lumen boundary. The line profile plot showed clear delineation between the lumen and wall of the carotid artery phantom. As shown in the plot in Figure 2-B, CT number consistency was reached at a radial distance of 0.38 mm in the wall region of the deconvolved image, while blooming caused a decreasing

CT number in the original non-deconvolved image. The CT number of contrast-enhanced lumen remained unaltered before and after deconvolution as shown in Figures 2-A and 2-B.

The results from image deconvolution applied to a single slice PCCT image (25–120 keV) of left common carotid artery (increased VV artery) of the porcine model is shown in Figure 3. The mean CT numbers of the wall of increased VV artery and the control artery are plotted in Figure 3.

The deconvolved images showed reduced CT number of the wall in both the control and increased VV arteries. Also, the CT number difference between the control and increased VV arteries was enhanced at all the time points after image deconvolution, while this difference is less prominent in the original images. Most time points, (after T=2, 6s after the bolus injection) showed enhanced CT number difference between the arteries, confirming increased wall perfusion in the left carotid with increased vasa vasorum density. The third time point showed the peak CT number following the contrast bolus injection (after 6s) with a CT number difference of approximately 30 HU between the two artery walls after image deconvolution. The results from the animal experiments show that the changes in the spatial density of vasa vasorum density can be reliably measured in PCCT images using image deconvolution. Moreover, the CT number difference was enhanced for a longer period of time after the contrast injection (from T=1 to T=8), compared to the original non-deconvolved data where the CT number difference is relatively less pronounced after T=2.

CONCLUSIONS

This study demonstrated the reduction in the blooming effects and subsequent improvement in CT number accuracy of artery wall measurements enabled by image deconvolution applied to PCCT images. Improved CT number consistency in artery wall as a consequence of reduced blooming was demonstrated in a 3D printed carotid phantom without altering the CT number of the contrast-enhanced lumen. In a porcine model, the CT number difference between an artery with increased vasa vasorum density in the wall and a normal control artery was enhanced, thereby allowing the detection of the increased wall perfusion which can serve as a maker of early atherosclerosis.

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Figure 1.

Image deconvolution applied to the 3D printed carotid phantom PCCT image at 25–120 keV. Note the sharp edges at lumen and the outer wall after deconvolution (right). Display window $[-400, 300]$ HU, scale bar = 3.81 mm. See Figure 2 for plot of the line profiles before and after image deconvolution.

Figure 2. Line profile plot showing lumen-wall delineation after image deconvolution (A) and mean CT number at different radial distances from the lumen boundary (B). After image deconvolution, CT number consistency of the wall was achieved after a radial distance of 0.38 mm.

Figure 3. Image deconvolution applied to PCCT image (25–120 keV) of increased VV artery with iodine in artery lumen. A noticeable reduction in blooming can be seen in the image after deconvolution (right) which also showed better structural delineation.

Figure 4.

Left: The mean CT number of the artery wall plotted for 10 time points involving the right carotid with normal VV density serving as the control (CNTL) and left carotid with induced proliferation of the vasa vasorum (IPVV) before image deconvolution. **Right:** The mean CT numbers of the artery wall plotted after deconvolution. The overall CT number has been reduced due to a reduction in blooming artifact. Note the increased difference in CT numbers between the increased VV and control curves, confirming the proliferation of vasa vasorum in the left carotid.