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Removal of arterial wall calcifications in CT angiography by local subtraction

Marcel van Straten^{a)}

Department of Medical Physics, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands

Henk W. Venema

Department of Medical Physics, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands, and Department of Radiology, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands

Geert J. Streekstra

Department of Medical Physics, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands

Jim A. Reekers and Gerard J. den Heeten

Department of Radiology, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands

Cornelis A. Grimbergen

Department of Medical Physics, Academic Medical Center, P.O. Box 22660 1100 DD, Amsterdam, The Netherlands

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CT Angiography (CTA) is an established technique for the minimally invasive imaging of arteries. The technique of maximum intensity projection (MIP) is often used to get a comprehensive overview of the vascular anatomy. On a MIP, however, arterial wall calcifications may hinder the visualization of the arterial lumen. These calcifications are in direct contact with the contrast-enhanced blood, which makes removal difficult. We present a local subtraction method for the automatic removal of these calcifications. In our approach a second CT scan has to be made, prior to contrast injection. The calcifications in both scans are registered prior to subtraction to compensate for displacements in between the two scans. Local subtraction results are compared with results obtained by thresholding. The method was tested in a phantom and with data from four patients. The phantom represented an artery with different types of stenosis. Data were used from patients for which CTA of the renal arteries was performed. For two patients the electrocardiogram (ECG) was recorded during the CTA examination, making retrospective cardiac gated reconstructions possible. Both the phantom and the patient study showed that the local subtraction method is capable of removing calcifications and visualizing the residual lumen. In the patient study it appeared that some artifacts remained for higher pitch values. We conclude that the local subtraction method is less subjective and more accurate than thresholding. Best results are obtained by use of a small pitch, at the expense of the volume covered during a single breath hold. © 2003 American Association of Physicists in Medicine. [DOI: 10.1118/1.1567271]

Key words: calcification, CT angiography, digital subtraction, partial volume effect, thresholding

I. INTRODUCTION

CT Angiography (CTA) has become an established technique for minimally invasive imaging of arteries in different parts of the human body.^{1,2} In order to get a comprehensive overview of the vascular anatomy the three dimensional (3D) visualization technique of maximum intensity projection (MIP) is used regularly. To be able to observe the arteries in a MIP image all high density structures have to be removed from the volume of interest prior to the application of the projection procedure.³ Especially arterial wall calcifications can be a serious problem in the visualization of the real lumen. This problem has been reported in a number of CTA studies concerning the renal arteries,⁴⁻⁶ the coronary arteries⁷ and the carotid arteries.⁸⁻¹⁰ Calcifications can also

complicate the use of volume rendering (VR). To differentiate calcifications from contrast-enhanced blood at VR one has to choose a transfer function with different opacity and/or color values for contrast-enhanced blood and calcification.¹¹ The choice for a particular transfer function strongly influences the visualization of the lumen and thus the grading of the stenosis.

Different methods exist for the removal of high density structures. Structures that are well separated from the arteries of interest can be removed manually by using a cutting tool on a graphical workstation. This can be done in multiple slices simultaneously if a viewing angle can be chosen in such a way that structures to be removed do not overproject the arteries of interest on a MIP. In case of overprojection in

all orientations the cutting has to be done for one slice at a time, which is very time consuming. In case the high density structures are contiguous to the arteries, which is the case for calcifications, this procedure is virtually impossible. Other methods like thresholding, region growing and masking¹² can be used to identify and remove high density structures. These methods have in common that the CT value of each voxel is either replaced by an arbitrary low value or left unaltered. Unfortunately, contrast-enhanced blood in the immediate vicinity of arterial wall calcifications has a relatively high CT value due to the nonzero width of the point spread function (PSF). Consequently choosing the parameters of the methods mentioned above so that all voxels belonging to high density structures are removed, without including contrast enhanced blood, is difficult and subjective. Therefore these methods are less suitable for the removal of arterial wall calcifications.

In this article we report on a method for the automatic removal of arterial wall calcifications. A technique that is commonly used in conventional angiography, but less so in CTA, is subtraction: a nonenhanced image is subtracted from the contrast-enhanced image in order to remove all common structures. Our method, local subtraction, is based on this principle. Due to the inevitable displacements of the patient in between the two scans the images have to be registered prior to subtraction. The method was tested in a phantom and with data from four patients. The phantom consisted of cylinders with two different types of stenosis. The patient data originated from a study in which patients underwent a CTA examination of the renal arteries. For two patients the electrocardiogram (ECG) was also recorded during data acquisition. This allowed us to make cardiac gated reconstructions retrospectively. Results of local subtraction are compared with results obtained by thresholding.

II. THEORY AND METHODS

A. Image formation

In a CT scan all structures are imaged with blurred boundaries, due to the finite width of the PSF. The width of the PSF depends on the construction of the CT scanner, a number of scanning parameters and the reconstruction kernel. We approximate the PSF by a 3D Gaussian:

$$\begin{aligned} \text{PSF}(\mathbf{x}, \boldsymbol{\sigma}) &\approx G(\mathbf{x}, \boldsymbol{\sigma}) \\ &= \frac{1}{(2\pi)^{3/2} \sigma_x \sigma_y \sigma_z} e^{-1/2(x^2/\sigma_x^2 + y^2/\sigma_y^2 + z^2/\sigma_z^2)}, \end{aligned} \quad (1)$$

with $\boldsymbol{\sigma}=(\sigma_x, \sigma_y, \sigma_z)$ the standard deviation of the Gaussian in three orthogonal directions and $\mathbf{x}=(x, y, z, 1)^T$ the homogeneous coordinate¹³ in the 3D image. The use of homogeneous coordinates facilitates the notation of the transformations, as will become clear in the next section.

The nonenhanced image $B(\mathbf{x}, \boldsymbol{\sigma})$ and contrast-enhanced image $C(\mathbf{x}, \boldsymbol{\sigma})$ are the result of the convolution of the object functions $B_O(\mathbf{x})$ and $C_O(\mathbf{x})$ with $\text{PSF}(\mathbf{x}, \boldsymbol{\sigma})$:

$$B(\mathbf{x}, \boldsymbol{\sigma}) = B_O(\mathbf{x}) * \text{PSF}(\mathbf{x}, \boldsymbol{\sigma}), \quad (2)$$

$$C(\mathbf{x}, \boldsymbol{\sigma}) = C_O(\mathbf{x}) * \text{PSF}(\mathbf{x}, \boldsymbol{\sigma}). \quad (3)$$

The object functions represent the distribution of the true CT values in 3D space. In the following the functions $B(\mathbf{x}, \boldsymbol{\sigma})$ and $C(\mathbf{x}, \boldsymbol{\sigma})$ are considered for discrete voxel coordinates $\mathbf{x}$ only.

We consider three types of material which are relevant to the removal of hyperdense structures: the hyperdense structure itself, blood (nonenhanced or contrast-enhanced) and the surrounding tissue. The mean CT value of hyperdense structures is relatively high. That of the tissue is relatively low. In the nonenhanced situation the CT values of blood and surrounding tissue are approximately equal. In the contrast-enhanced situation the CT value of blood clearly exceeds the CT values of the surrounding tissue.

B. Calcification removal by local subtraction

The principle of calcification removal is illustrated with an example which is shown in Fig. 1. In this example the arterial lumen is represented by a cylinder with a diameter of 6.0 mm. The cylinder contains a dense structure, contiguous to the perimeter, which has to be removed.

In the upper part of Fig. 1 the object functions $B_O(\mathbf{x})$ and $C_O(\mathbf{x})$ are shown. In the middle part the functions $B(\mathbf{x}, \boldsymbol{\sigma})$ and $C(\mathbf{x}, \boldsymbol{\sigma})$ are shown. The latter pair of functions is obtained by convolution of the object functions with an isotropic Gaussian PSF with $\sigma=0.6$ mm. The CT values across a horizontal line through the center of the cylinder are shown in the lower part of Fig. 1.

The local subtraction procedure only pertains to the voxels of the hyperdense structure and the peripheral zone of this structure. The inclusion of this peripheral zone is necessary to subtract all voxels whose CT values are altered due to the presence of the hyperdense structure. The collection of voxels belonging to the dense structure and its peripheral zone is denoted by R . This collection is determined by thresholding, followed by binary dilation. Details are given in Sec. III B and Sec. III C for the phantom and patient study, respectively. The boundaries of R are indicated in Fig. 1 by dotted vertical lines. The voxels in R in the nonenhanced situation are subtracted from the corresponding voxels in the contrast-enhanced situation. Because subtraction of only a part of an image produces a discontinuity in the CT values, the subtracted CT values are scaled and an offset value is added in order to restore the continuity:

$$L(\mathbf{x}, \boldsymbol{\sigma}) = \begin{cases} (C(\mathbf{x}, \boldsymbol{\sigma}) - B(\mathbf{x}, \boldsymbol{\sigma})) \cdot \gamma + \delta, & \forall \mathbf{x} \in R, \\ C(\mathbf{x}, \boldsymbol{\sigma}), & \forall \mathbf{x} \notin R. \end{cases} \quad (4)$$

The scale factor γ is given by

$$\gamma = \frac{\text{CT}_{\text{contrast-enhanced lumen}} - \text{CT}_{\text{surrounding tissue}}}{\text{CT}_{\text{contrast-enhanced lumen}} - \text{CT}_{\text{nonenhanced lumen}}}, \quad (5)$$

with $\text{CT}_{\text{material}}$ the mean CT value of a specific material. The offset value δ is equal to $\text{CT}_{\text{surrounding tissue}}$. The result of local subtraction for the data of Fig. 1 is shown in Fig. 2(a).

FIG. 1. Model of an artery consisting of a cylinder with a stenosis caused by a hyperdense structure which covers 25% of the cylinder wall. (a) Nonenhanced situation. (b) Contrast-enhanced situation. Top: Object functions $B_O(x)$ and $C_O(x)$. Middle: $B(x, \sigma)$ and $C(x, \sigma)$, obtained by convolution of the object functions with a Gaussian PSF. Bottom: CT values across a horizontal line through the center of the cylinder. The dotted vertical lines indicate the position of the boundaries of the peripheral zone of the hyperdense structure used in the local subtraction method.

In practice the hyperdense structures have to be registered prior to local subtraction to compensate for displacements in between the two scans. All structures are registered individually because they can move in relation to each other. Each structure R_i in $B(x, \sigma)$ is registered by applying a 3D rigid transformation T_i , consisting of translation and rotation. The transformation T_i is determined by minimizing the cost function;

$$F(T_i) = \sum_{\mathbf{x} \in R_i} (B(\mathbf{x}, \sigma) - C(T_i \mathbf{x}, \sigma))^2. \quad (6)$$

Linear interpolation is used for the calculation of the value of $C(T_i \mathbf{x}, \sigma)$ for positions $T_i \mathbf{x}$ not corresponding with voxel coordinates. The minimization is performed by the downhill simplex method.¹⁴ This method needs initial values for the six transformation parameters, and six characteristic variations around these initial values. The initial values are determined by registration of the cluster of hyperdense structures

FIG. 2. Result of removal of the hyperdense structure in Fig. 1(b). (a) Removal by local subtraction. (b) Removal by thresholding. Top: cross-sectional image (with a zero width PSF) of the model with the dense structure replaced by material with a CT value of surrounding tissue. Middle: cross-sectional images of the model after removal of the hyperdense structure. Bottom: CT values across a horizontal line through the center of the cylinder in the top image (dashed lines) and middle images (solid lines).

as a whole (see also Sec. III C). As characteristic variation 1 mm was taken for the translations and 1 degree for the rotations.

The registration procedure described above may introduce a bias on the transformation T_i . The main reason for this bias is that the CT value of the lumen is increased in the contrast-enhanced situation. Due to the nonzero width of the PSF this affects the shape of the dense structure, which is immediately adjacent to the blood. In the example of Figs. 1 and 2 no registration is needed, as the dense structure is on the same location in B_O and C_O . The minimum value of the cost function, however, is found for a small shift in the x -direction, $\Delta x = 0.14$ mm. When this shift is implemented, an artifact appears in the locally subtracted image [see Fig. 6(c), cross-sectional image 2]. In the following we discuss a method to minimize this artifact.

The magnitude of the bias depends on the width of the PSF. In order to obtain a small bias the PSF should be as narrow as possible. A narrower PSF, however, corresponds

with an increase in image noise, which is generally undesirable in CTA of the abdominal arteries. To be able to use the PSF that is chosen for the clinical protocol, we compute T_i for several values of $|\sigma| > 0$, extrapolate the components of T_i to the situation $\sigma = 0$, and apply the unbiased transformation to the image with the desired PSF. The in-plane width of the PSF (σ_x and σ_y) can be decreased by choosing another reconstruction kernel. The width of the PSF can be increased in all three directions by convolving the data with a Gaussian kernel $G(\mathbf{x}, \sigma_G)$:

$$B(\mathbf{x}, \sigma') = B(\mathbf{x}, \sigma) * G(\mathbf{x}, \sigma_G), \quad (7)$$

$$C(\mathbf{x}, \sigma') = C(\mathbf{x}, \sigma) * G(\mathbf{x}, \sigma_G), \quad (8)$$

with the standard deviation σ' of the resulting PSF given by

$$\sigma'_j = \sqrt{(\sigma_j)^2 + (\sigma_{G,j})^2}, \quad j = x, y, z. \quad (9)$$

The bias correction method described above was tested in both the phantom and patient study. The shape and dimensions of the stenosis in the phantom study are equal to that of the example used in this section.

C. Calcification removal by thresholding

Results of the local subtraction method, described in the previous section, will be compared with results obtained by thresholding. Removal of a high density structure by thresholding is given by

$$M(\mathbf{x}, \sigma) = \begin{cases} C(\mathbf{x}, \sigma), & \text{if } C(\mathbf{x}, \sigma) < \tau, \\ \varepsilon, & \text{if } C(\mathbf{x}, \sigma) \geq \tau, \end{cases} \quad (10)$$

with τ the threshold value and ε an arbitrary low CT value. The value of τ should be as low as possible in order to remove as many as possible voxels belonging to a hyperdense structure. The CT value of the contrast-enhanced lumen is the lower limit for τ . In practice the value of τ must be higher, due to the presence of noise in the CT image. The result of thresholding the data of Fig. 1(b) is shown in Fig. 2(b).

III. EXPERIMENTS

A. Measurement of the point spread function

A small gold sphere (diameter 0.22 mm), embedded in a synthetic material was used to measure the point spread function of the CT scanner. The PSF was measured for the scan parameters used in the phantom and patient study. Reconstructions were made with kernel A (smooth) and kernel D (sharp) with a voxel size of $0.125 \times 0.125 \times 0.125 \text{ mm}^3$ (much smaller than the full width at half maximum of the PSF).

B. Phantom study

A phantom was designed to represent a renal artery with a diameter of 6.0 mm (see Fig. 3). It includes a hyperdense structure mimicking a calcified plaque all around the vessel wall, and a smaller structure which covers only 25% of the vessel wall. In the larger hyperdense structure a small low-

FIG. 3. Schematic view of the phantom with dimensions in millimeters. (a) Longitudinal section. (b) Cross-section. Three cylinders (white), which can be filled with water with iodine, represent arteries with a diameter of 3 mm (top) and 6 mm (center and bottom). The phantom is constructed from lucite (gray). The hyperdense structures in the center and bottom cylinder are made of PVC (black).

density cylinder is inserted, to mimic an occlusion. It also includes a cylinder with a diameter of 3.0 mm which represents the gold standard for the lumen within the first hyperdense structure. The phantom was constructed from lucite and polyvinylchloride (PVC), which are approximately tissue- and calcification-equivalent with respect to their attenuation coefficients (CT values approximately 130 and 950 HU, respectively).

The orientation of the phantom during the scans, denoted by the angle α between the cylinder axis and the direction of table movement, was varied. The phantom was scanned for three orientations: $\alpha = 0^\circ$, $\alpha = 45^\circ$ and $\alpha = 90^\circ$. For the nonenhanced scan the cylinders were filled with water with a low iodine concentration, resulting in a mean CT value of 140 HU. For the contrast-enhanced scan the cylinders were filled with water with a higher iodine concentration, resulting in a mean CT value of 410 HU. The increase of the attenuation coefficient between the two scans (270 HU) corresponded approximately with the difference between attenuation coefficients of nonenhanced and contrast-enhanced blood in the CTA examination of the renal arteries.

Spiral CT datasets were acquired with a CT scanner with four detector arrays (Mx8000; Philips, Eindhoven, The Netherlands). Both nonenhanced and contrast-enhanced scans were made with a collimation of $4 \times 1.0 \text{ mm}$, a scan time of 0.75 s per 360° rotation, 120 kV, 300 mAs per section. Two values for the pitch were used: 0.875 (table feed 4.7 mm/s) and 0.375 (table feed 2.0 mm/s). Nonionic contrast material (Iohexol; [Omnipaque]; Nycomed Amersham, Oslo, Norway; 300 mg of iodine per milliliter) was used. Reconstructions were made with a voxel size of $0.20 \times 0.20 \times 0.60 \text{ mm}^3$, 360° interpolation and reconstruction kernel A (smooth). Additional reconstructions were made with kernel D (sharp) for a determination of the bias on the registration parameters.

TABLE I. Scan parameters used for the four patients.

No.	Pitch	Rotation time (s/360°)	Heart rate (BPM)
1	0.875	0.75	-
2	0.875	0.75	-
3	0.425	0.50	55–60
4	0.450	0.50	45–60

The hyperdense structures were identified in the nonenhanced image by thresholding with $\tau=500$ HU. The extra region around each structure was obtained by binary dilatation with a spherical kernel with a radius of 2.5 mm.

C. Patient study

Data were used from four patients for which CTA of the renal arteries was performed. First a low dose localization scan was made to visualize the kidneys and calcifications, if any. In case calcifications were present in or near the renal arteries, a regular dose nonenhanced spiral CT scan was made. To keep the additional radiation dose as low as possible, this scan was made only of the region containing the calcifications. Finally the CT angiographic examination was performed. The contrast material was injected in a cubital vein at a rate of 3 ml/s for a period equal to the total scan time. Scanning was started after a delay time that was measured with a delay test. The study was approved by our institutional review board and informed consent was obtained from all patients. Table I gives an overview of the pitch and rotation time for these patients. For two patients (numbers 1 and 2) the scan parameters were the same as in the phantom study. For the other two patients (numbers 3 and 4) the electrocardiogram (ECG) was recorded simultaneously and the scan parameters were adapted to be able to make retrospective cardiac gated (tagged) reconstructions.^{15,16} The tagged reconstructions were made to investigate the amount of motion of calcifications during data acquisition and the influence on the local subtraction result. As the CT scans of patients 3 and 4 were made with a lower pitch; the untagged reconstructions of these scans allowed us to investigate the influence of the pitch on the image quality. For these last two patients the rotation time was decreased to 0.5 s per 360° rotation. Reconstructions were made with a voxel size of $0.49 \times 0.49 \times 0.70$ mm³ while the other reconstruction parameters were the same as in the phantom study. A conventional angiogram was made from patient 2, six months after the CTA examination, so that results of these two modalities could be compared with respect to the visualization of the lumen at the location of the (removed) calcifications.

Each cluster of 6-connected voxels with $B(\mathbf{x}, \sigma) > \tau$ was labeled as a separate structure and the extra region around each structure was obtained by binary dilatation with a spherical kernel with a radius of 2.5 mm. The threshold value τ was chosen to be 50 to 70 HU above the mean value of the nonenhanced blood in order to avoid the inclusion of voxels representing blood. The range of 50–70 HU is in the order of three to four times the standard deviation of the CT values in a region of interest in the nonenhanced blood. In the regis-

tration procedure the set of clusters was first registered as a whole to avoid convergence problems during the individual registration of each cluster, and next each cluster was registered separately.

Image processing was performed on a personal computer equipped with a 650 MHz processor (Pentium III processor; Intel, USA). The processing time for the removal of the calcifications by local subtraction and thresholding depended on the number of clusters found. For a typical number of clusters (5–10) the processing time was approximately 15 minutes.

IV. RESULTS

A. Point spread function

The PSF is analyzed in two directions: the z -direction (direction of table movement) and the radial direction (in the x, y -plane). The measured PSFs can be very well approximated by Gaussians. The standard deviations of the fitted Gaussians are $\sigma_A = (0.60, 0.60, 0.60)$ mm for reconstruction kernel A and $\sigma_D = (0.34, 0.34, 0.60)$ mm for reconstruction kernel D. The full width at half maximum (FWHM) of the PSF in the z -direction, which is often used as a measure for the effective slice thickness, is for both reconstruction kernels equal to $\sigma_z \sqrt{8 \ln 2} \approx 1.41$ mm.

B. Phantom study

The results of local subtraction and thresholding were evaluated on both cross-sectional and MIP images. Figure 4 shows images and profiles along a line for both the nonenhanced and contrast-enhanced CT scan (with $\alpha=0^\circ$ and pitch 0.875) of the middle cylinder of Fig. 3. The contrast-enhanced CT scan is also depicted after removal of the high density structure by both local subtraction and thresholding. In this case no registration bias is present because of the circular symmetry of the high density structure and the contrast-enhanced lumen. The CT values along the center line in the subtracted image are in exact agreement with the CT values along the same line in the top cylinder in Fig. 3 (not shown). Figure 5 shows results of local subtraction for all combinations of cylinder orientation and pitch value. Both phantom orientation and pitch value do influence the quality of the result. Best results are obtained for $\alpha=0^\circ$. For $\alpha=45^\circ$ and $\alpha=90^\circ$ the results for low pitch are better.

Figure 6 shows the result of removal of the asymmetric high density structure in the bottom cylinder of Fig. 3. Note that the physical properties of this stenosis correspond with the data used in Fig. 1. For this stenosis the transformation T_i found during registration depends on the (effective) width of the PSF. This dependency is shown in Fig. 6(a) for the translation Δx . The value of Δx was determined for three values of the standard deviation of the PSF: $\sigma_x = \sigma_y = 0.34$ mm (reconstruction kernel D), $\sigma_x = \sigma_y = 0.60$ mm (reconstruction kernel A) and $\sigma'_x = \sigma'_y = 0.90$ mm (reconstruction kernel A and convolution with a Gaussian kernel

FIG. 4. MIP images (top left), cross-sections (top right) and profiles along the centerline (bottom) for CT scans of the 6 mm cylinder (Fig. 3, middle) with $\alpha=0^\circ$ and pitch 0.875. (a) Nonenhanced and (b) contrast-enhanced CT scan. (c) Result of removal by local subtraction. (d) Result of removal by thresholding (with $\tau=450$ HU). The lines in the images indicate the location of the plotted profile. The dashed vertical lines in the intensity profiles represent the physical boundaries of the high density structure.

with $\sigma_G=0.67$ mm). In the three situations the value of σ_z was 0.60 mm, 0.60 mm and 0.90 mm, respectively. Because of the symmetrical shape of the hyperdense structure in the z -direction, however, the exact value of σ_z does not influence the results. The measurements show that it is reasonable to assume a linear relationship between σ_x and Δx . By extrapolation to $\sigma_x=0$ mm the bias on Δx is removed [Figs.

6(b) and 6(c)]. The influence of σ on the translation in the y - and z -direction is very small, because of the symmetry of the high density structure in these directions, as is the influence on the rotation parameters (not shown).

FIG. 6. Reduction of the bias on the transformation. (a) Extrapolation of the value of the translation in the x -direction. (b) Profiles of the result of local subtraction for four values of Δx . (c) Cross-sectional images of the result of local subtraction for four values of Δx . The numbers correspond with the registration results after using images that are reconstructed with kernel D (1), with kernel A (2) and with kernel A plus a Gaussian blur (3). The results denoted by (0) correspond with the unbiased transformation.

FIG. 5. MIP images after local subtraction for different values of the pitch and for different orientations α of the phantom.

FIG. 7. Transverse images and profiles along a line of CT scans of the aorta of patient 1 with arterial wall calcifications. (a) Registered nonenhanced and (b) contrast-enhanced CT scan. The calcification on the left side of the aorta is used for the registration. The result after removal of the arterial wall calcifications by (c) local subtraction and (d) thresholding. The lines in the images indicate the location of the plotted profile. The dashed lines represent the profile from figure (b).

C. Patient study

The mean CT value of the nonenhanced blood was approximately 50 HU. The mean CT value of the contrast-enhanced blood varied between approximately 200 and 300 HU. The maximum CT value of the calcifications varied between 1100 and 1600 HU for images reconstructed with kernel A.

Cross-sectional images and profiles along a line of the nonenhanced and contrast-enhanced CT scan of patient 1 are shown in Figs. 7(a) and 7(b); respectively. The results of the removal of calcifications in the aortic wall are shown in Fig. 7. For the removal by thresholding the lowest threshold that was feasible in this patient ($\tau=280$ HU, only slightly above 200 HU, the mean CT value of the contrast-enhanced lumen) does not remove all voxels that may obscure the lumen on a MIP. The local subtraction method does remove the calcification in such a way that a clean MIP can be made (MIP not shown).

MIP and multiplanar reformation (MPR) images of the contrast-enhanced CT scan of patient 2 are shown in Fig. 8. On the MPR image, made after removal by local subtraction, some artifacts remain at the location of the removed calcifications. We tried the bias reduction method that we successfully applied in the phantom study, but the resulting images were virtually the same (not shown). Evidently this artifact must have another cause. We return to this point in the discussion. On the MPR image, made after removal by thresholding (with $\tau=300$ HU), the width of the lumen (mean CT value of 230 HU) appears to be reduced significantly. It is also clearly visible that the high density voxels at the transi-

FIG. 8. MIP images (left) and MPR images (right) of the renal artery with arterial wall calcifications (patient 2). (a) Original contrast-enhanced image. (b) Calcification removed by local subtraction. (c) Calcification removed by thresholding. The lines in the MIP images indicate the location of the MPR planes.

FIG. 9. Conventional digital subtraction angiogram of patient 2.

tion from calcification to surrounding tissue are not removed. The conventional DSA study (made six months after the CTA examination) shows that the diameter of the lumen is not reduced at the location of the calcifications (Fig. 9).

For patient 3 tagged reconstructions are made for 10 equidistant phases of the cardiac cycle. For a calcification at the origin of the renal artery the location of the center of mass for each phase is calculated. The maximum shift, relative to the location of the calcification at the start of the cardiac cycle, is approximately 1.5 mm. The result of local subtraction and thresholding is shown in Fig. 10. Additionally a typical example of local subtraction with tagged reconstructions is depicted. The tagged reconstructions are made at 30% of the cardiac cycle.

Cross-sectional images of the nonenhanced and contrast-enhanced CT scan of patient 4 are shown in Figs. 11(a) and 11(b), respectively. In this case untagged reconstructions are used. In Fig. 11 the result of calcification removal is shown.

V. DISCUSSION AND CONCLUSIONS

In this article we dealt with the removal of arterial wall calcifications when making MIP images in case these calcifications hinder the diagnosis of a possible stenosis. Although in the diagnostic process radiologists always include the source images and will use MPR images as well, the processed MIP images can be of great value: source images often do not have the correct orientation for a proper judgement of the calcified areas and with the MPR technique there is a strong investigator dependence on the choice of the orientation of the MPR plane. Furthermore, referring clinicians often have less sophisticated possibilities for visualization and may feel uncomfortable when making clinical decisions using MIP images in which the lumen of the arteries cannot be depicted.

The most obvious method for the removal of arterial wall calcifications is thresholding.^{8,17,18} The selection of the value of the threshold, however, is critical. In the literature rather *ad-hoc* choices have been made.⁹ Thresholding is only effective in removing all voxels associated with the calcifications if it is performed with a low value of the threshold. The drawback, however, is that voxels that actually represent blood, but have CT values above the threshold value due to the nonzero width of the PSF, or noise, are also removed. In small vessels this means a relatively large reduction of the

FIG. 10. MIP images (left) and MPR images (right) of renal artery with arterial wall calcifications (patient 3). (a) Original contrast-enhanced image (untagged). (b) Calcification removed by local subtraction, using untagged reconstructions. (c) Calcification removed by thresholding, using untagged reconstructions. (d) Calcification removed by local subtraction, using tagged reconstructions. The lines in the MIP images indicate the location of the MPR planes.

lumen and consequently an overestimation of the grade of stenosis. We have used thresholding as well and, as expected, this method did not remove the calcifications satisfactory, both in the phantom and in the patient study.

The local subtraction method uses information from an additional CT scan made prior to the injection of the contrast agent. Because of the inevitable movement of calcifications relative to each other in between the two scans, conventional global subtraction cannot be applied. A possible approach would be the application of an elastic registration method. This approach is much more involved, however, and any global subtraction method has the drawback of a deterioration of the signal-to-noise ratio.¹² In our method the calcifi-

FIG. 11. Transverse images and profiles along a line of CT scans of the aorta of patient 4 with arterial wall calcifications. (a) Registered nonenhanced and (b) contrast-enhanced CT scan. Result after removal of the arterial wall calcifications by (c) local subtraction and (d) thresholding. The lines in the images indicate the location of the plotted profile. The dashed lines represent the profile from (b).

cations in the nonenhanced image are subtracted from the corresponding calcifications in the contrast-enhanced image, leaving the rest of the image unaltered. The continuity of the CT values is restored by scaling of the subtracted CT values and by the addition of an offset value. This procedure works perfectly if the CT values of the tissue in the immediate neighborhood of the calcifications are constant; otherwise small discontinuities will remain. In the four patients considered here, the latter appeared not to be the case.

In the phantom study the high density structures were removed correctly by local subtraction. In case of the asymmetrical high density structure, the theoretically predicted bias in the transformation parameters for the registration was indeed present, and the correction method, based on the extrapolation to the situation of a zero width PSF, removed this bias. Although satisfactory results were obtained for all scans in the phantom study, the best results were obtained for the low pitch scans. This can be explained by the fact that spiral CT inevitably is accompanied by slight interpolation artifacts.¹⁹ The artifacts are mainly present at structures that change rapidly in the z -direction. In most cases these artifacts are hardly noticeable in the images themselves, but they may show up in the subtracted images because the artifacts in most situations are not exactly the same in the two CT scans. The magnitude of the artifacts depends on the scan parameters and reconstruction method. Especially the pitch is an important parameter.²⁰ It thus appears that the use of a low pitch is an important factor in obtaining local subtraction results with the highest quality.

In the patient study the local subtraction method yielded results of varying quality. The quality of the subtraction depends primarily on the quality of the registration of the cal-

cifications. The steep transition in CT value from calcification to surrounding tissue, and to blood, is the reason that local subtraction results in relatively large differences in CT values, when even a minimal mismatch of a calcification is present. These CT values may be erroneously interpreted as contrast-enhanced blood. A mismatch can be caused by different reasons, which can be classified in two categories: incorrect registration of the calcifications and differences in the depicted shape of the calcifications in the two scans. An incorrect registration can be caused by the bias in the transformation parameters. As noted above, this bias was successfully removed in the phantom study. In the patient study, however, the bias correction method did not improve the quality of the results noticeably. We note that the bias reduction method used in this article can be improved upon, as the full range of σ was only applied in the x - and y -direction, while in the z -direction the standard deviation for the smallest scale equals that of the medium scale. For the smallest scale the value can actually be reduced by using a deconvolution technique.²¹ We do not think that for the patient study the extension of the bias reduction method would lead to substantial improvement, as in this case the bias reduction method hardly had any effect on the quality of the calcification removal. This was even the case when the bias could be expected to be present in the x -, y -plane only, as is the case for instance in calcifications in the aortic wall.

We are of the opinion that slight changes in depicted shape of the calcifications in the CT images are the main reason for the imperfections in the removal of the calcifications. The influence of the interpolation artifacts on the depicted shape, as discussed above for the phantom study, can be expected to be present in the clinical data as well. How-

ever, in the patient study the shape of calcifications is also influenced by the motion of calcifications during data acquisition. In an attempt to reduce the influence of motion during data acquisition, two patients were scanned while simultaneously recording the ECG, which allows for retrospective cardiac gated reconstructions. These two scans revealed a displacement of the calcifications during the cardiac cycle of approximately 1.5 mm, caused by vessel movements and pulsation due to the heart beat. With regard to the ECG correlated reconstructions, the subtraction results appeared to depend on the cardiac phase chosen for reconstruction. However, no link could be found between the movement of a calcification during the cardiac cycle, and the quality of the result of local subtraction. Use of these gated reconstructions showed no improvement over the local subtraction obtained without gating. An explanation is that additional artifacts are induced by the gated reconstruction technique, for this technique uses only a selection of the transmission measurements during the selected phase of the cardiac cycle. The best results were actually obtained when using the untaged reconstructions for these two patients. These results (with pitch 0.425 and 0.450) appear to be better than the results obtained for the other two patients (with pitch 0.875). This is in accordance with the phantom study where better local subtraction results were obtained at the lower pitch values.

We cannot exclude the possibility that motion artifacts were a problem in the registration for the two patients in which CT scans were made without the option of cardiac gating as for these patients a higher pitch was used than in the gated studies. The use of a high pitch may engender a different influence of the motion of calcifications on the image reproducibility. Further research is needed to reveal the relation between scan parameters and motion blurring on the one hand and the relation between scan parameters and reconstruction artifacts on the other.

We conclude that the local subtraction method, compared to the thresholding method, is less subjective and more accurate by taking the effects of the intrinsic blur of the CT scanner into account. Best results are obtained by use of a small pitch, at the expense of the volume covered during a single breath hold. With the introduction of helical cone-beam CT scanners with a larger coverage in the *z*-direction this problem will be less severe and we expect our method to be even more valuable.

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^a)Electronic mail: m.vanstraten@amc.uva.nl

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