Image processing in vascular computed tomography
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Removal of bone in CT angiography by multiscale matched mask bone elimination

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ABSTRACT

For clear visualization of vessels in CT angiography (CTA) images of the head and neck using maximum intensity projection (MIP) or volume rendering (VR) bone has to be removed. In the past we presented a fully automatic method to mask the bone (matched mask bone elimination, MMBE) for this purpose. A drawback is that vessels adjacent to bone may be partly masked as well. We propose a modification, multiscale MMBE, which reduces this problem by using images at two scales: a higher resolution than usual is used for image processing, and a lower resolution to which the processed images are transformed for use in the diagnostic process. A higher in-plane resolution is obtained by the use of a sharper reconstruction kernel. The out-of-plane resolution is improved by deconvolution or by scanning with narrower collimation. The quality of the mask that is used to remove bone is improved by using images at both scales. After masking, the desired resolution for the normal clinical use of the images is obtained by blurring with Gaussian kernels of appropriate widths. Both methods (multiscale and original) were compared in a phantom study and with clinical CTA data sets. With the multiscale approach the width of the strip of soft tissue adjacent to the bone that is masked can be reduced from 1.0 mm to 0.2 mm without reducing the quality of the bone removal. The clinical examples show that vessels adjacent to bone are less affected and therefore better visible. Images processed with multiscale MMBE have a slightly higher noise level or slightly reduced resolution compared with images processed by the original method and the reconstruction and processing time is also somewhat increased. Nevertheless multiscale MMBE offers a way to remove bone automatically from CT angiography images without affecting the integrity of the blood vessels. The overall image quality of MIP or VR images is substantially improved relative to images processed with the original MMBE method.
I. INTRODUCTION

Computed tomography angiography (CTA) makes it possible to visualize the arteries and veins in a minimally invasive way. Visualization is often performed by means of maximum intensity projection (MIP) or volume rendering (VR) so that the contrast-enhanced vessels can be examined easily. In a CTA data set the maximum value along a line in the direction of projection often belongs to a bone structure, which obscures the vessels in MIP images. In VR images bone may also affect the visualization of the vessels because of an overlap of the CT values of bone and the contrast agent in the vessels. Therefore the voxels that represent bone have to be removed from the data set prior to visualization.1–3

Manual removal of bone from CTA images can be very time consuming or virtually impossible when arteries are in close proximity of bone. Automated methods are based on image processing techniques, such as region-growing and thresholding.4–8 Unfortunately these methods are time-consuming as well, and they are subjective because they require user-interaction. Because it is difficult to find the exact border between bone and contrast-enhanced vessels, the results of these techniques are not optimal, i.e. the bone removal is often incomplete or part of the vasculature is removed.

Matched mask bone elimination (MMBE) is a fully automatic technique and has proven to be a robust method for bone removal.9–12 For CTA examinations with the MMBE method two scans are made: a low dose scan (a quarter of the dose of the CTA scan) without contrast injection and a normal CTA scan. After registration of the two scans, the nonenhanced scan is converted into a binary mask, which is then used to remove the bone in the CTA scan. A problem that arises when masking a bone in a CT image, is that a thin strip of adjacent tissue is masked as well due to the non-zero width of the point spread function (PSF) (Fig. 1). When arteries or veins are in close proximity of bone this may affect their integrity as they are eroded by the mask. CTA examinations are reconstructed in the clinical routine with a rather smooth reconstruction kernel in order to achieve an acceptable noise level. Therefore the resulting in-plane point spread function is broad and a relatively broad strip adjacent to the bone is masked.
The width of the PSF in the \( z \)-direction (the slice width), which is the direction of the table movement is also not negligible. This will also result in a relatively broad strip adjacent to the bone that is masked in the \( z \)-direction.

In this paper we present an alternative approach, multiscale MMBE, that addresses the problem of erosion of the vessels. In multiscale MMBE CT images are used that have a higher resolution than usual in image processing and masking. The quality of the mask is improved by combining the information from two versions of the nonenhanced scan, a high resolution and a low resolution image, i.e. at two scales. Because of the higher resolution the bone and the vessels will have less overlap, and therefore the vessels will be less affected in the masking process. Images with less noise, that are desired for the diagnostic process, are obtained in the final step by blurring the processed images.

We compare the performance of the multiscale and the original MMBE method in a phantom study and examples of CTA scans of patients.

II. MULTISCALE MMBE

A flow-diagram of the multiscale MMBE method is shown in Fig. 2 The method consists of the following steps:

1a. Reconstruction of the nonenhanced scan and the CTA scan with a sharper reconstruction filter than the one that is normally used for CTA examinations.

1b. Increase the resolution in the \( z \)-direction, either by deconvolution\(^{13,14}\) when...
the CT scan can not be made with a smaller slice width, or by scanning with a smaller slice width when this is possible.

2. Registration of the high resolution nonenhanced and CTA images.

3-6. Construction of the mask. The procedure is more involved than in the original MMBE method since the increased noise level in the high resolution images has to be taken into account.

7. Masking of the bone voxels.

8. Blurring of the sharp images to obtain images that can be used in the diagnostic process.

9. Production of MIP or VR images.

The original MMBE method uses smooth images with a relatively large slice width and consists of steps 2, 3, 7 and 9. The different steps of multiscale MMBE are discussed below.

Fig. 2
Flow diagram of the multiscale MMBE method. Below the images the symbolic notations are denoted that are used in the different stages of multiscale MMBE (see Sec. II.A to II.D).
II. A. LOW AND HIGH RESOLUTION CT IMAGES (STEP 1)

In a CT scan all structures are imaged with blurred boundaries with an unsharpness that can be quantified by the point spread function (PSF). We approximate the PSF by a three dimensional (3D) Gaussian:\cite{15}

$$PSF(x,\sigma) \approx G(x,\sigma) \equiv \frac{1}{(2\pi)^{\frac{3}{2}}\sigma_x\sigma_y\sigma_z} e^{-\frac{1}{2} \left( \frac{x^2 + y^2}{\sigma_{xy}^2} + \frac{z^2}{\sigma_z^2} \right)}$$

with $x=(x,y,z)$, $x$ and $y$ the in-plane coordinates and $z$ the coordinate in the direction of the table movement, and $\sigma=(\sigma_x, \sigma_y, \sigma_z)$ the standard deviation of the Gaussian in three directions. Because of the in-plane symmetry $\sigma_x = \sigma_y = \sigma_{xy}$. The standard deviation $\sigma$ is in this study also indicated with the term scale.

The reconstructed nonenhanced images, $B(x,\sigma)$ and the contrast-enhanced images, $C(x,\sigma)$ are the result of the convolution of the corresponding object functions $B_0(x)$ and $C_0(x)$ with the PSF:\

$$B(x,\sigma) = B_0(x) * PSF(x,\sigma)$$

$$C(x,\sigma) = C_0(x) * PSF(x,\sigma)$$

The object functions represent the distribution of the true CT values in 3D space. In practice $B(x,\sigma)$ and $C(x,\sigma)$ are available for discrete voxel coordinates only; they are denoted as $B(\sigma)$ and $C(\sigma)$.

The in-plane resolution and the resolution in the $z$-direction depend on different factors. The in-plane resolution depends on factors related to the data acquisition, such as the focal spot size, detector size, geometric factors, integration time in the read-out of the detectors, and on the convolution kernel used in the reconstruction process, which may vary from smooth (low resolution) to sharp (high resolution). The standard deviation of the in-plane PSF of reconstructions made with the smooth and the sharp reconstruction kernel are denoted by $\sigma_{xy}^{low}$ and $\sigma_{xy}^{high}$, respectively.

The resolution in the $z$-direction is determined by the choice of the slice width. With older scanners, the possibilities for the reduction of the slice width are limited,
because with these scanners a high \( z \)-resolution can only be obtained by using narrow collimation which prolongs the scan time. When narrow collimation is not used some improvement in the resolution can be obtained with deconvolution. With state-of-the-art multi-slice CT scanners it is possible to make scans with a very small slice width in a short scan time, and thus the high \( z \)-resolution can be obtained straightforward. In the present study we used both scanning with a relatively large slice width followed by deconvolution, and scanning with a narrow slice width, to obtain images with a higher resolution in the \( z \)-direction. The standard deviation of the PSF in the \( z \)-direction for the images with the lower and the higher resolution are denoted by \( \sigma_{z}^{\text{low}} \) and \( \sigma_{z}^{\text{high}} \), respectively.

After image processing the smooth images that are required for the final visualization are obtained from the sharp images by means of blurring with a 3D Gaussian kernel with standard deviation \( \sigma_{\text{blur}} \). This procedure does not exactly restore the combination of in-plane resolution and noise of the reconstructions obtained with the smooth kernel. The reason is that the Gaussian kernel is only an approximation to the kernel that is required for the transformation between sharp and smooth images, and because of reasons of noise aliasing. However, the Gaussian approximation is reasonable, as will be shown in this paper. In the \( z \)-direction no such problems are expected.

**II.B. REGISTRATION (STEP 2)**

The nonenhanced images and the CTA images are registered to compensate for patient movements between the two scans. The same method is used as in the original MMBE method.

**II.C. CONSTRUCTION OF THE BONE MASK (STEP 3-6)**

The basic step of the construction of the bone mask consists of thresholding of the registered nonenhanced images, \( B_{\text{regis}} (\sigma^{\text{high}}) \). Thresholding alone, however, produces a bone mask of unacceptable quality. This is caused by the large number of voxels that are erroneously incorporated in the mask due to the noise in the image. To overcome this problem an additional smoothed image \( B_{\text{regis}} (\sigma^{\text{low}}) \) is made, which is obtained by convolution of the sharp image \( B_{\text{regis}} (\sigma^{\text{high}}) \) with a
Gaussian kernel with standard deviation $\sigma_{\text{blur}}$. This image is also thresholded.

For both images the same threshold $\tau$ is used, which results in two binary masks, the sharp and the smooth mask (Fig. 3b and Fig. 3c). A preliminary mask is obtained by taking all voxels that the smooth and sharp mask have in common (Fig. 3d). This step removes the isolated voxels from the sharp mask that were incorporated due to noise.

In the construction of the preliminary mask also some voxels corresponding to small bone structures are removed because they are below the threshold $\tau$ in the smooth version of the nonenhanced images. Most of these bone structures, e.g. in the inner ear, the nasal cavity and the paranasal sinuses, are surrounded by air, which will cause their intensity to decrease substantially in the blurring process. These voxels are identified by making an additional binary mask (the high-decrease mask) by taking the difference of the sharp image and the smooth image, and thresholding this difference image with threshold $\delta$. The voxels that are present in both the high-decrease mask and the sharp mask (Fig. 3e) are added to the preliminary mask to obtain the final bone mask (Fig. 3f).

In the original MMBE method the bone mask is dilated slightly to improve the quality of the bone removal. Dilation will remove structures close to the mask that have intensities below $\tau$, but still may hinder visualization of vessels. In multiscale MMBE this step is less important because the final blurring (see next paragraph) reduces the intensity of any residual bone edge to a considerable amount. Dilation of the mask was therefore investigated as an optional step.
**II.D. MASKING AND BLURRING (STEP 7 AND 8)**

In the masking step the CT values of the voxels in the sharp CTA scan, \( C(\sigma^{\text{high}}) \) that are present in the bone mask are replaced by a value \( \zeta \). Next Gaussian blurring is applied with \( \sigma^{\text{blur}} \) to obtain the clinically desired noise level. After the Gaussian blurring the images can be visualized by MIP or VR.

**II.E. PARAMETERS**

Multiscale MMBE has seven parameters, three scale parameters (with two components) and four parameters used in the image processing: (1) \( \sigma^{\text{low}} \), the scale of the images used in clinical practice; (2) \( \sigma^{\text{high}} \), the scale chosen for the processing of the images, (3) \( \sigma^{\text{blur}} \), the scale that is used to convert the high-resolution into the low-resolution images, (4) threshold \( \tau \) that is used for the preliminary version of
the mask, (5) threshold $\delta$ that is used for refinement of the mask, (6) the number of voxels that is used in the (optional) dilation step and (7) the CT value $\zeta$ that is given to the masked voxels.

We consider the first parameter $\sigma_{\text{low}}$ to be fixed, as this is the resolution used in clinical practice in the radiology department in our hospital. The third parameter, $\sigma_{\text{blur}}$, can to a first approximation be derived from $\sigma_{\text{low}}$ and $\sigma_{\text{high}}$, as will be discussed in the next section. Therefore five main parameters remain that have to be chosen, $\sigma_{\text{high}}$ (with two components, $\sigma_{\text{high}}^1$ and $\sigma_{\text{high}}^2$), $\tau$, $\delta$, the number of voxels dilatation, and $\zeta$. The choice of these parameters is explained in the next section.

III. EXPERIMENTS

III.A. CT SCANS

CT scans were made with a CT scanner with four detector arrays (Mx8000 Quad; Philips Medical Systems; Best; The Netherlands). All scans were made with a tube voltage of 120 kV, head mode, high resolution, 0.75 sec per 360 degrees rotation and a pitch of 0.875 (table feed 4.7 mm/s), unless indicated otherwise. The collimation used was either 4 x 1 mm (nominal slice width 1.3 mm) or 2 x 0.5 mm (nominal slice width 0.6 mm). All reconstructions were made with a field-of-view of 150 mm and a 512 x 512 matrix, resulting in a pixel size of 0.293 x 0.293 mm$^2$. For use in multiscale MMBE reconstructions were made with kernel D (sharp), which is the sharpest reconstruction kernel available, and a slice increment of 0.1 mm. A small slice increment was chosen because this is preferable when deconvolution is used. The scans were also reconstructed with kernel B (smooth) and a slice increment of 0.5 mm, which is the routine setting for reconstruction of CTA scans in our hospital, for processing with the original MMBE method. The Fourier transforms of kernels B and D for the head mode and high resolution are shown in Fig. 4. All reconstructions were made using an iterative beam hardening correction (UltraImage option).
III.B. REDUCTION OF THE SLICE WIDTH

1. COLLIMATION 4 X 1 MM

The resolution in the $z$-direction of the images acquired with a collimation of 4 x 1 mm was increased (from $\sigma_z^{\text{low}}$ to $\sigma_z^{\text{high}}$) with the aid of constrained iterative deconvolution.\textsuperscript{13,14} The value of $\sigma_z^{\text{high}}$ that is obtained is determined by the choice of the parameter $\sigma_z^{\text{dev}}$, with

$$\sigma_z^{\text{high}} = \sqrt{(\sigma_z^{\text{low}})^2 - (\sigma_z^{\text{dev}})^2} \quad (4)$$

The deconvolution consist of a number of steps, each step consisting of improvement of the longitudinal resolution, followed by blurring of the improved
image with $\sigma_z^{dcv}$. This procedure is continued until the difference between the blurred improved image and the original image is smaller than a predetermined threshold.

In the deconvolution procedure we choose $\sigma_z^{dcv} = 0.25$ mm. In this way only a limited reduction of the width of the PSF was obtained. This choice was made, however, in order to keep the introduction of edge ringing artifacts in the deconvolved images that are inevitable in a deconvolution procedure to a minimum.\(^\text{14}\)

2. COLLIMATION 2 X 0.5 MM
The images acquired with 2 x 0.5 mm already have the desired higher resolution in the z-direction ($\sigma_z^{high}$). After image processing and masking the images are blurred to a lower resolution ($\sigma_z^{low}$). We choose this lower resolution to be the same as the original z-resolution of the images acquired with a collimation of 4 x 1 mm for reasons of comparison.

III.C. MEASUREMENT OF THE POINT SPREAD FUNCTION
The standard deviation of the in-plane PSF was determined by making a scan of a cylindrical water-filled phantom with a diameter of 20 cm containing three steel wires (diameter 0.15 mm) that were aligned in the z-direction. The scan was made with 250 effective mAs. Reconstructions were made with kernels B and D and a pixel size of 0.1 x 0.1 mm\(^2\). By fitting the convolution of a disk (diameter 0.15 mm) with a two dimensional (2D) Gaussian with a free parameter $\sigma_{xy}$ to the reconstructed image data, the standard deviation of the in-plane PSF was estimated. This was done for each of the wires in five different images separately and $\overline{\sigma}_{xy}$ was obtained as the average value.

The standard deviation of the PSF in the z-direction was determined from scans of an edge of a cube made of polyvinyl chloride (PVC) with a CT number of approximately 1100 HU, surrounded by water. The scans were made with 250 effective mAs. The reconstructions were made with kernel D and a voxel size of 0.293 x 0.293 x 0.1 mm\(^3\). From a region of 80 by 50 by 70 voxels containing the edge of the cube, 11 sagittal images of 80 by 70 pixels were used to estimate the standard deviation of the PSF in the z-direction by fitting a convolution of a
Gaussian and a step edge to the image data. The estimate of \( \bar{\sigma}_z \) was obtained as the average standard deviation of these eleven images.

### III.D. DETERMINATION OF \( \sigma^{\text{blur}} \)

In case the PSFs are 3D Gaussian functions, the components of \( \sigma^{\text{blur}} \), which is used to blur the sharp images to the desired smooth resolution, (see Sec. II.A) are given by:

\[
\sigma_{xy}^{\text{blur}} = \sqrt{\left(\sigma_{xy}^{\text{low}}\right)^2 - \left(\sigma_{xy}^{\text{high}}\right)^2}
\]

(5)

and

\[
\sigma_z^{\text{blur}} = \sqrt{\left(\sigma_z^{\text{low}}\right)^2 - \left(\sigma_z^{\text{high}}\right)^2}
\]

(6)

Because the PSFs are only approximately Gaussian and because of the noise aliasing to be expected, we determined experimentally the value of \( \sigma^{\text{blur}} \) that has to be used to obtain images with the desired resolution and noise level. The resolution was measured for images blurred with a number of values of \( \sigma^{\text{blur}} \) as described in the previous section.

The noise level was quantified by measuring the standard deviation of the noise \( (SD_n) \) in a homogeneous region of interest, containing water, with an area of approximately 3 cm\(^2\). Each noise measurement was performed in three images and averaged.

### III.E. CHOICE OF THE IMAGE PROCESSING PARAMETERS

For the threshold for inclusion in the preliminary mask we used \( \tau = 150 \) HU. This value is the same as the threshold that was found optimal by Van Straten et al.\textsuperscript{11} for the original MMBE method. For the threshold for inclusion in the high-decrease mask it appeared that \( \delta = 250 \) HU produced satisfactory results. For the CT value of the masked bone voxels we choose \( \zeta = 20 \) HU, which is approximately the value of brain tissue.

Multiscale MMBE was used both without and with dilation. Two different
amounts of dilation were tried, a kernel of four in-plane adjacent voxels, and a kernel of ten adjacent (eight in-plane and two out-of-plane) voxels. These kernels refer to a in-plane voxel size of 0.293 mm and an out-of-plane voxel size of 0.5 mm. The last kernel is also used in the original MMBE method.

In multiscale MMBE all image-processing was performed with a slice increment of 0.1 mm. One should expect that in this case dilation with five 0.1 mm voxels in the $z$-direction should be applied to obtain the same amount of dilation as with one 0.5 mm voxel. However in the case of 0.1 mm voxels the position of the border of the mask is slightly shifted as well, and it appeared that the same amount of dilation in the $z$-direction was obtained using three 0.1 mm voxels.

For the parameter values of the original MMBE method we choose the values that were found optimal by Van Straten et al.\textsuperscript{11} apart from one small change: the value of $V_{\text{min}}$, the minimum volume of a connected object to be masked, was set to zero. This choice was made in order not to bias the original MMBE method negatively as further explained in the discussion.

### III.F. PHANTOM STUDY

The influence of the bone removal method on the width of the strip of low density material adjacent to bone that is masked was investigated with a phantom. The phantom consists of a small block of PVC of approximately $3 \times 4 \times 4$ cm$^3$, with a CT value of approximately 1100 HU. In this block three 5.0 mm diameter holes were drilled with angles of ninety and forty-five degrees (Fig. 5a). In these holes three cylinders made of polyoxymethylene (POM), with a diameter of 5.0 mm and a length of 40 mm, and with a CT value of approximately 300 HU can be inserted (Fig. 5b). These POM cylinders can also be placed in a holder of polymethylmethacrylate (PMMA) with the same three directions of the cylinders as in the cube of PVC (Fig. 5c). The block of PVC and the PMMA holder were placed in a water-filled cylinder made of PMMA, with an outer diameter of 5 cm. The phantom was aligned with one of the holes (or one of the cylinders) in the $z$-direction of the scanner. To obtain a noise level comparable to that in clinical scans, the cylinder containing the phantom was scanned in a ring made of PMMA with an outer diameter of 15 cm.

The three configurations of this phantom represent the different situations of
vessels in a CTA examination: nonenhanced in bone (a), contrast-enhanced in bone (b), and contrast-enhanced in tissue (as reference) (c).

Scans of the phantom were made using both 4 x 1 mm and 2 x 0.5 mm collimation. In both cases scans were made with 65 effective mAs for configuration (a), and 250 effective mAs for configuration (b) and (c). For the scans made with 2 x 0.5 mm collimation, the images that were used in the original MMBE method were obtained by Gaussian blurring of the thin slices in the z-direction to obtain the slice width of scans made with 4 x 1 mm collimation. Multiscale MMBE was applied using the scans of configuration (a) and (b).

The width of the masked strip of low-density material adjacent to bone was estimated by measuring the diameter of the cylinders in the processed images and the reference images. Cross sectional images of the cylinders were obtained by reslicing the 3D dataset. The direction of each cylinder was found by fitting a straight line to the center line points obtained by an automatic method for center line detection. The diameter of the cylinders in two orthogonal directions was measured by fitting an ellipse convolved with two Gaussian functions with different standard deviations in the x- and y-direction to the cross-sectional images of the cylinders. This was done in 30 to 35 different cross sectional images of each cylinder with a spacing of approximately 1 mm, and the average was determined.

Fig. 5
A diagram of the three configurations of the phantom: (a) the PVC cube with water-filled holes, (b) the PVC cube filled with three POM cylinders and (c) the POM cylinders in a small PMMA holder. This represents three different situations of vessels in a CTA examination: nonenhanced in bone (a), contrast-enhanced in bone (b), and contrast-enhanced in tissue (as reference) (c).
**III.G. EXAMPLES OF PATIENT STUDIES**

Examples of two CTA examinations of the circle of Willis that were processed with multiscale MMBE, and with the original MMBE method for comparison, are presented. In one examination 4 x 1 mm collimation was used, in the other 2 x 0.5 mm collimation.

In the first patient (collimation 4 x 1 mm) 65 effective mAs was used for the nonenhanced scan, 250 effective mAs for the CTA scan and a tilt of -16°. Images were reconstructed with a nominal slice width of 1.3 mm, and processed using deconvolution.

In the second patient (collimation 2 x 0.5 mm) 100 effective mAs was use for the nonenhanced scan, 175 effective mAs for the CTA scan, a tilt of -12.5° and a pitch of 1.25. Images were reconstructed with a nominal slice width of 0.6 mm. In this case the final resolution in the z-direction ($\sigma_z$) was chosen such that the same slice width was obtained as in the first example, as already mentioned in section IIIb. Images with this same slice width were also processed with multiscale MMBE with deconvolution and the original MMBE method for comparison.

In both examples the quality of bone removal was quantified by measuring the standard deviation in a region of interest in the MIP image of 50 x 70 pixels and 42 x 42 pixels respectively in the MIP image where (almost) no vessels are present. For purposes of comparison all MIP-images were made using a slice increment of 0.5 mm.

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**IV. RESULTS**

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**IV.A. MEASUREMENT OF THE POINT SPREAD FUNCTION**

The values of the estimated $\sigma_{xy}$ for the smooth and sharp kernel are shown in Table I. Also shown are the values of $\sigma_z$ for 4 x 1 mm collimation before and after deconvolution and for 2 x 0.5 mm collimation.
Table I
Measured values of $\sigma_{xy}$ for the kernel B and D, $\sigma_z$ for 4 x 1 mm collimation before and after deconvolution, and $\sigma_z$ for 2 x 0.5 mm collimation

<table>
<thead>
<tr>
<th></th>
<th>symbol</th>
<th>$\sigma$ (mm)</th>
<th>fwhm (mm)*</th>
</tr>
</thead>
<tbody>
<tr>
<td>in-plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>kernel B (smooth)</td>
<td>$\sigma_{xy}^{low}$</td>
<td>0.431</td>
<td>1.015</td>
</tr>
<tr>
<td>kernel D (sharp)</td>
<td>$\sigma_{xy}^{high}$</td>
<td>0.271</td>
<td>0.638</td>
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<td>z-direction</td>
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<td>4 x 1 collimation</td>
<td>$\sigma_z^{low}$</td>
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<td>4 x 1 collimation after deconvolution</td>
<td>$\sigma_z^{high}$</td>
<td>0.498</td>
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<td>2 x 0.5 collimation</td>
<td>$\sigma_z^{high}$</td>
<td>0.301</td>
<td>0.708</td>
</tr>
</tbody>
</table>

* fwhm (full width at half maximum) = $\sigma \times \sqrt{8 \ln 2}$

IV.B. DETERMINATION OF $\sigma_{xy}^{blur}$

The relation between $\sigma_{xy}$ and the standard deviation of the noise ($SD_n$) after Gaussian blurring with for a number of values of $\sigma_{xy}^{blur}$ is shown in Fig. 6. Using the values in Table I and formula (5) one calculates that for Gaussian PSFs blurring with a $\sigma_{xy}^{blur} = 0.335$ mm should restore the resolution of the smooth images (kernel B). As expected slight deviations occur: the resolution is restored with $\sigma_{xy}^{blur} = 0.404$ mm, the noise level with $\sigma_{xy}^{blur} = 0.451$ mm. Restoring the resolution results in a noise penalty of 18 %; restoring the noise level results in a 9 % increase in width of the PSF. We choose for this last option, for reasons mentioned in the discussion.
Fig. 6
Relation of \( \sigma_{xy} \) and the standard deviation of the noise (SDn) after Gaussian blurring with a number of values of \( \sigma_{xy}^{blur} \). The two reconstruction filters: sharp, and smooth are denoted with a \( \bullet \) and \( \blacktriangle \) marker, respectively. The resulting \( \sigma_{xy} \) and SDn after blurring with different values of \( \sigma_{xy}^{blur} \) are denoted with + markers. The standard deviation of the measurements are shown as horizontal or vertical bars.

For the blurring in the \( z \)-direction we found that the calculated values using Table I and formula (6), and the experimentally determined values of \( \sigma_{z}^{blur} \) are in good agreement. For the deconvolved scans made with 4 x 1 mm collimation, blurring in the \( z \)-direction with the calculated value \( \sigma_{z}^{blur} = 0.254 \) mm produced images with a slice width within 1 % of the desired slice width. Blurring with this calculated value of \( \sigma_{z}^{blur} = 0.254 \) mm did also restore the noise level within 1 %.

For scans made with 2 x 0.5 mm collimation blurring with the calculated value of \( \sigma_{z}^{blur} = 0.471 \) mm produced images within 1 % of the desired slice width. The noise level of these images was 4 % higher than the noise level of images acquired with 4 x 1 mm. This is a satisfactory agreement considering all the approximations involved.
IV.C. PHANTOM STUDY

The effect of the bone removal method on the profile of the CT values on a line through the center of a cross-section of the cylinder is shown in Fig. 7. Each line is an average of lines in 35 images with a spacing of 1 mm. From this figure the gain of multiscale MMBE is clear. The profile after application of multiscale MMBE resembles the ‘true’ profile (cylinder in water) much closer than when the original MMBE method is used. When dilation is performed the gain is somewhat reduced, but is still present. We note that one should expect the CT values in the center of the cylinder to be the same in the reference situation and after processing with MMBE. The small differences in CT-value were probably caused by residual errors in the beam-hardening correction, or by slight calibration errors of the CT-scanner.

![Fig. 7](image)

The estimated diameters of the cylinders with different orientations in the scans processed with multiscale MMBE and original MMBE are listed in Table II. For multiscale MMBE also dilation of the mask with four and ten voxels was used. For comparison the estimated diameters in the reference situation, the 5.0 mm cylinders in water, are also shown.
The widths of the strip adjacent to the bone that are masked with the different methods are shown in Table III. These values were obtained as the half of the difference of the reference diameter and the diameter in the processed images.

The estimated values in the $xy$-plane and in the plane with an angle of 45° are virtually the same for scans with a collimation of 4 x 1 mm and 2 x 0.5 mm. For 2 x 0.5 mm collimation the values for the cylinder with an angle of 90° with the $z$-axis are also comparable to those determined in the other orientations. For 4 x 1 mm collimation some anomalies are present in the measured diameters in the processed images in this last orientation. The estimated diameter in the $x$-direction appears to be somewhat low, and in the $z$-direction too high. Moreover in the measurements in the $z$-direction a trend was present, the diameter in the center of the cylinder being in the order of 0.15 to 0.25 mm higher than the average value and in the periphery up to 0.5 mm lower. These anomalies are probably due to interpolation artifacts in the $z$-direction at the high-contrast interface of the cylinder and the PVC surrounding, which appear to be more severe for 4 x 1 mm than for 2 x 0.5 mm collimation.

The standard deviations of the measurements were in the order of 0.05 mm for all reference diameters. In the processed images the standard deviation was in the order of 0.1 - 0.15 mm, with exception of the measurements in the $z$-direction in scans with a collimation of 4 x 1 mm. In this case the standard deviation was 0.24 mm due to the trend mentioned above.
### Table II
The estimated diameters of the POM cylinders (diameter 5.0 mm) after application of multiscale MMBE, multiscale MMBE with 4 and 10 voxels dilation and original MMBE (with 10 voxels dilation). For comparison the estimated reference diameters are shown in the last column.

<table>
<thead>
<tr>
<th>angle with the z-axis</th>
<th>diameter direction</th>
<th>4 x 1 collimation estimated diameter (mm)</th>
<th>2 x 0.5 collimation estimated diameter (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>4 x 1 collimation</td>
<td>2 x 0.5 collimation</td>
</tr>
<tr>
<td></td>
<td></td>
<td>multiscale MMBE 4 dilation</td>
<td>multiscale MMBE 4 dilation</td>
</tr>
<tr>
<td></td>
<td></td>
<td>multiscale MMBE 10 dilation</td>
<td>multiscale MMBE 10 dilation</td>
</tr>
<tr>
<td></td>
<td></td>
<td>original MMBE</td>
<td>original MMBE</td>
</tr>
<tr>
<td></td>
<td></td>
<td>reference situation</td>
<td>reference situation</td>
</tr>
<tr>
<td>0°</td>
<td>x</td>
<td>4.47</td>
<td>3.94</td>
</tr>
<tr>
<td>0°</td>
<td>y</td>
<td>4.59</td>
<td>3.94</td>
</tr>
<tr>
<td>45°</td>
<td>x</td>
<td>4.48</td>
<td>3.89</td>
</tr>
<tr>
<td>45°</td>
<td>y</td>
<td>4.49</td>
<td>3.87</td>
</tr>
<tr>
<td>45°</td>
<td>y/z</td>
<td>4.49</td>
<td>3.87</td>
</tr>
<tr>
<td>90°</td>
<td>x</td>
<td>4.24</td>
<td>3.70</td>
</tr>
<tr>
<td>90°</td>
<td>y</td>
<td>4.58</td>
<td>3.88</td>
</tr>
<tr>
<td>90°</td>
<td>y/z</td>
<td>4.56</td>
<td>3.98</td>
</tr>
<tr>
<td>90°</td>
<td>z</td>
<td>4.86</td>
<td>4.36</td>
</tr>
<tr>
<td>90°</td>
<td>z</td>
<td>4.60</td>
<td>4.15</td>
</tr>
</tbody>
</table>
### Table III
The calculated width of the masked strip adjacent to bone in multiscale MMBE, multiscale MMBE with 4 and 10 dilation, and in the original MMBE method.

<table>
<thead>
<tr>
<th>4 x 1 collimation masked strip adjacent to bone (mm)</th>
<th>angle with the z-axis</th>
<th>diameter direction</th>
<th>multiscale MMBE</th>
<th>multiscale MMBE 4 dilation</th>
<th>multiscale MMBE 10 dilation</th>
<th>original MMBE</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>x</td>
<td>0.21</td>
<td>0.51</td>
<td>0.61</td>
<td>0.96</td>
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</tr>
<tr>
<td></td>
<td>y</td>
<td>0.23</td>
<td>0.55</td>
<td>0.67</td>
<td>0.99</td>
<td></td>
</tr>
<tr>
<td>45°</td>
<td>x</td>
<td>0.26</td>
<td>0.56</td>
<td>0.66</td>
<td>0.93</td>
<td></td>
</tr>
<tr>
<td></td>
<td>y/z</td>
<td>0.24</td>
<td>0.55</td>
<td>0.69</td>
<td>0.97</td>
<td></td>
</tr>
<tr>
<td>90°</td>
<td>x</td>
<td>0.35</td>
<td>0.62</td>
<td>0.65</td>
<td>1.24</td>
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</tr>
<tr>
<td></td>
<td>z</td>
<td>0.15</td>
<td>0.40</td>
<td>0.62</td>
<td>1.01</td>
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</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>2 x 0.5 collimation masked strip adjacent to bone (mm)</th>
<th>angle with the z-axis</th>
<th>diameter direction</th>
<th>multiscale MMBE</th>
<th>multiscale MMBE 4 dilation</th>
<th>multiscale MMBE 10 dilation</th>
<th>original MMBE</th>
</tr>
</thead>
<tbody>
<tr>
<td>0°</td>
<td>x</td>
<td>0.20</td>
<td>0.54</td>
<td>0.65</td>
<td>0.96</td>
<td></td>
</tr>
<tr>
<td></td>
<td>y</td>
<td>0.24</td>
<td>0.59</td>
<td>0.71</td>
<td>0.98</td>
<td></td>
</tr>
<tr>
<td>45°</td>
<td>x</td>
<td>0.18</td>
<td>0.54</td>
<td>0.64</td>
<td>0.88</td>
<td></td>
</tr>
<tr>
<td></td>
<td>y/z</td>
<td>0.23</td>
<td>0.52</td>
<td>0.65</td>
<td>0.99</td>
<td></td>
</tr>
<tr>
<td>90°</td>
<td>x</td>
<td>0.25</td>
<td>0.50</td>
<td>0.57</td>
<td>0.81</td>
<td></td>
</tr>
<tr>
<td></td>
<td>z</td>
<td>0.22</td>
<td>0.45</td>
<td>0.62</td>
<td>1.04</td>
<td></td>
</tr>
</tbody>
</table>

### IV.D. PATIENT EXAMPLES

The difference between multiscale MMBE and the original MMBE method is illustrated in Fig. 8 on axial images of the first patient. The first column shows a sharp CTA image masked using multiscale MMBE without and with dilation of the bone mask with four adjacent in-plane voxels. The middle column shows the same images after blurring to the desired resolution. The last column of Fig. 8 shows the image masked using the original MMBE method without dilation and with dilation of the bone mask with 8 in-plane and 2 out-of-plane voxels. In the images processed with multiscale MMBE (middle row) less erosion of vessels is visible in proximity of bone in comparison with original MMBE (bottom right), which confirms the results of the phantom study.
In Fig. 9 MIP images of the CTA examination of the circle of Willis of the first patient are shown (4 x 1 mm collimation). The quality of the bone removal of multiscale without dilation (Fig. 9a) is comparable with the quality of the original MMBE method (Fig. 9d). In both images some small bone remnants remain in the inner ear and the nasal cavity. Bone removal is more complete when dilation is used (Fig. 9 b and c) The standard deviation in the ROIs (see Fig. 9) where the skull base is masked are listed in Table IV.
Table IV
Quality of bone removal in two patients quantified by the standard deviation in a ROI without vessels after multiscale MMBE, multiscale MMBE with 4 and 10 voxels dilation, and the original MMBE method.

<table>
<thead>
<tr>
<th>Patient 1 (Fig. 9; 4 x 1 mm collimation)</th>
<th>SD ROI (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>multiscale MMBE</td>
<td>27</td>
</tr>
<tr>
<td>multiscale MMBE (4 dilation)</td>
<td>20</td>
</tr>
<tr>
<td>multiscale MMBE (10 dilation)</td>
<td>19</td>
</tr>
<tr>
<td>original MMBE</td>
<td>27</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Patient 2 (Fig. 11; 2 x 0.5 mm collimation)</th>
<th>SD ROI (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>multiscale MMBE (thin slices, no dilation)</td>
<td>26</td>
</tr>
<tr>
<td>multiscale MMBE (thick slices, no dilation)</td>
<td>33</td>
</tr>
<tr>
<td>original MMBE</td>
<td>36</td>
</tr>
</tbody>
</table>

With multiscale MMBE less erosion of the vessels is present when they are close to bone. This can be observed in more detail in Fig. 10 where a region of interest (ROI) is shown of the location where the internal carotid artery passes through the skull base. Fig. 10 b and c show the trade-off of the quality of bone removal and the erosion of the vessels when dilation is used.

Fig. 9
Coronal MIP images (130 x 58 mm²) of a CTA examination of the circle of Willis. Window center: 50 HU. Window width: 300 HU. Bone was removed in (a) with multiscale MMBE without dilation, (b) with four voxels dilation and (c) with 10 voxels dilation. Image (d) was processed with the original MMBE method. The ROI where the SD is measured to quantify the bone removal is shown in the lower left.
MIP images of the second patient (2 x 0.5 mm) are shown in Fig. 11. The result of multiscale MMBE using the original images is shown in Fig. 11a and b. The result of multiscale MMBE using the images with the same slice width as in the 4 x 1 mm scan, and deconvolution, is shown in Fig. 11c and d. In both approaches no dilation was used. The results of original MMBE method is shown in Fig. 11e and f. The standard deviation in the ROIs at the lower right in Fig. 11(a, c and e) are listed in Table IV.

In this example the main difference between multiscale MMBE using the thin slices (2 x 0.5 mm collimation) and the thicker slices followed by deconvolution, appears to be the more effective bone removal of the first method. No substantial differences can be seen in the integrity of the vessels. In both respects the multiscale method performs better than the original MMBE method.
Fig. 11
Sagital (80 x 40 mm²) and coronal (50 x 40 mm²) MIP images of a CTA examination of the circle of Willis. Window center: 200 HU. Window width: 400 HU. Bone was removed with (a and b) multiscale MMBE with narrow collimation, (c and d) multiscale MMBE with deconvolution and (e and f) classic MMBE. The ROI where the SD is measured to quantify the bone removal (in the region of the nasal cavity) is shown in the lower right (a, c and e).
V. DISCUSSION

In this study multiscale MMBE is presented which is a modification of a method for automated bone removal from CTA scans. The main advantage of the multiscale approach over the original MMBE method is that the width of the strip of soft tissue adjacent to bone that is sacrificed in the masking process is substantially reduced, while the quality of the bone removal is retained or even improved.

In the phantom study the thickness of the strip of low density material adjacent to the bone, that is masked erroneously, was measured. The width of this strip is a measure for the maximum erosion of blood vessels adjacent to bone when MMBE is applied. In the original MMBE method the width of this strip is approximately 1.0 mm. With multiscale MMBE, the width of this strip was only 0.2 mm. When the mask was dilated with four voxels the width of the strip increased to 0.5 mm, dilation with 10 voxels increased it to 0.6 - 0.7 mm.

The gain that is obtained by using the sharper reconstruction kernel alone, can be observed by comparing the results of multiscale MMBE and the original method with the same amount of dilation of 10 voxels. In this case the erroneously masked strip is reduced from 1.0 to 0.6 - 0.7 mm. An additional advantage in multiscale MMBE is, however, that less dilation of the mask is needed than in the original MMBE method for the same quality of bone removal. This is because the final blurring step, required to reduce the noise to an acceptable level, has the positive side effect that the residual unmasked bone edges are reduced to a considerable extent.

From patient examples it is clear that removal of bone with multiscale MMBE is as effective as in the original MMBE, even when multiscale MMBE is used without dilation. With dilation of four voxels the quality of bone removal was even better than in the original MMBE. With the parameter settings used in the present study, one could therefore use either no dilation or dilation with four voxels to obtain a good trade-off between the quality of the bone removal and the preservation of the vessels, depending on the preference of the observer.

In our study we changed the setting of one parameter of the original MMBE method that we used for purposes of comparison. This parameter is the minimum volume of connected voxels, $V_{min}$, that is included in the mask. This parameter was
introduced to eliminate spurious voxels due to noise in the mask. The use of the original value of $V_{\text{min}}$ of 40 mm$^3$ appeared to be counterproductive, however, as small bones or bone fragments remain unmasked, and noise did not appear to be a problem. Therefore in order not to bias the original MMBE method negatively, we decided not to use this criterion.

We compared two methods to improve the resolution in the $z$-direction: deconvolution in case the CTA examination was made with the usual collimation (4 x 1 mm), or scanning with narrower collimation (2 x 0.5 mm). Scanning with narrow collimation is more effective and thus preferable. With deconvolution the resolution can only be improved by a modest amount, otherwise edge ringing artifacts are introduced. Deconvolution has the further drawback that it is computational somewhat more expensive and that a small reconstruction increment has to be used, which prolongs the image reconstruction time. In the present study an increment of 0.1 mm was used, instead of an increment of 0.5 mm for 4 x 1 mm collimation. For a CT-scan with a length of 80 mm, for example, this means that an additional 640 images have to be reconstructed, which takes in the order of 10 tot 15 minutes for the CT-scanner used in the present study.

Scanning with sub-mm slice width, 2 x 0.5 mm instead of 4 x 1 mm collimation, to improve the $z$-resolution with the CT-scanner used in the present study, increases the scan time fourfold, all other things being equal. Therefore this last setting is not used routinely for CTA examinations. With state-of-the-art multislice CT-scanners, however, CTA examinations can be performed with sub-mm slice width (in the order of 0.5 to 0.7 mm) in a very short scan time. In this case scanning with very thin slices is a viable option, and processed images with a somewhat greater slice width and reduced noise level can be used in the final visualization step.

As mentioned in section II.E the value of four parameters have to be chosen for the processing of the high-resolution images: thresholds $\tau$ and $\delta$ that are used for the preliminary version and the refinement of the mask, respectively, the number of voxels that is used in the (optional) dilation step, and the value $\zeta$ that is given to the masked voxels. The optimal value $\tau$ for the main thresholding step depends primarily on the CT-values of soft tissue and of bone, respectively, and as these values are approximately constant in CTA examinations of the head and neck consistent results of the thresholding step are to be expected. We note that a higher value of $\tau$ has to be used when scans are made with a lower tube voltage,
as the CT-values of bone will then be systematically higher.\textsuperscript{10} It appeared in some test experiments that the choice of the threshold $\delta$ is not critical, which is to be expected as it is used in the refinement of a bone mask that has already acceptable quality. The choice of the number of voxels used in the (optional) dilation of the mask was discussed above. The value $\zeta$ that is given to the masked voxels, was formally introduced as a parameter; it is an obvious choice, however, as in the masking process the bone voxels are replaced by voxels representing soft tissue, and therefore we used $\zeta = 20$ HU which is approximately the CT-value of soft tissue in the brain and neck. We finally note that in this study we have used images at the two scales $\sigma^{\text{high}}$ and $\sigma^{\text{low}}$, in the construction of the mask. Strictly spoken the lower scale that is used in this step should also be considered as a parameter, as it can be choosen freely. Because the high and the low scale were already considerably different, which is the main point in using two scales, we considered this to be an issue of minor importance, and we have not done so.

In the present implementation the final blurring step is performed with a Gaussian kernel. This approximation appears to be adequate for the $z$-direction. For the in-plane situation, we have shown that when the high-resolution images are blurred with a Gaussian kernel, an acceptable approximation of the images reconstructed with a smooth kernel can be obtained. Because of the approximation involved and because of noise aliasing, restoration of the PSF increases the noise level slightly, and restoration of the noise level increases the width of the PSF slightly. We choose for this last option, as radiologists appear to be relatively insensitive for small variation in resolution when judging CTA images of the brain.

The use of a Gaussian kernel in the final blurring step is computational efficient, as the kernel is separable. An exact transformation of the sharp to smooth images in the spatial domain is also feasible,\textsuperscript{16} albeit at the cost of efficiency, as the separability of the kernel is then lost. We are currently investigating this possibility.

In addition to the approach of the present study other approaches to improve the image quality are also feasible. Recently, Van Straten et al.\textsuperscript{19} presented another modification, called soft MMBE, in which a combination of masking and local subtraction was applied. Application of subtraction at the bone-vessel interface does also reduce the erosion of vessels by the mask. Subtraction, however, inevitable leads to increase of noise in the subtracted region. It has the further drawback that the continuity needs to be restored between the original and the subtracted areas,
and that errors in the estimated offset value may show up as artifacts in the final images. Moreover, high-quality subtraction images are required, which can only be obtained when no significant interpolation artifacts are present. Therefore only scans can be used that are made at a low pitch, which is only feasible in CT-scanners with a large number of detector arrays.

Another approach is the local adjustment of the threshold value above which the bone is masked. By increasing the threshold value at the bone-vessel interface, less vessel will be masked. A drawback of this method is that the intensity profile of the vessel will still be influenced by the presence of the bone and will increase at the edges. When the original MMBE method is used this artifact is also present (see Fig. 7). In multiscale MMBE the intensity profile of the vessel decreases slowly at the edges due to smoothing step after the masking of the bone. This will resemble the true intensity profile of the vessel, only without the drawbacks of soft MMBE mentioned above.

A drawback of all masking methods in which an additional CT-scan is used is the additional radiation dose for the patient. This argument applies to multiscale MMBE, the original MMBE-method, and the above mentioned alternative approaches as well. The increase in radiation dose with multiscale MMBE is only 25%, however, and the benefit of the masking of bone in the clinical routine is considerable.

We conclude that the multiscale MMBE method is a substantial improvement over the original MMBE method and offers a way to remove bone from a CT angiography scan in a fully automatic and accurate way.
REFERENCES


