A New Filtering Algorithm for Medical Magnetic Resonance and Computer Tomography Images

Erich Stoll, Christian Stern, Peter Stucki, and Simon Wildermuth

Inner views of tubular structures based on computer tomography (CT) and magnetic resonance (MRI) data sets may be created by virtual endoscopy. After a preliminary segmentation procedure for selecting the organ to be represented, the virtual endoscopy is a new postprocessing technique using surface or volume rendering of the data sets. In the case of surface rendering, the segmentation is based on a grey level thresholding technique. To avoid artifacts owing to the noise created in the imaging process, and to restore spurious resolution degradations, a robust Wiener filter was applied. This filter working in Fourier space approximates the noise spectrum by a simple function that is proportional to the square root of the signal amplitude. Thus, only points with tiny amplitudes consisting mostly of noise are suppressed. Further artifacts are avoided by the correct selection of the threshold range. Afterwards, the lumen and the inner walls of the tubular structures are well represented and allow one to distinguish between harmless fluctuations and medically significant structures. Copyright © 1999 by W.B. Saunders Company

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VIRTUAL ENDOSCOPY VIEWING (VIE) is a new post processing technique of computer tomography (CT) and magnetic resonance (MR) data sets that allows one to create inner views of tubular structures.¹ Endoscopic perspectives can be obtained by surface or volume rendering of the data sets, but a preliminary segmentation procedure is required to select the organ to be represented. In the case of surface rendering, the segmentation is based on a thresholding technique, through which the range of Hounsfield densities, in case of CT, and a range of grey levels, in case of MR, is selected to represent the voxels pertaining to the structure. The incorrect selection of this range alters the representation of endoscopic findings and can also cause the appearance of artifacts, involving the lumen and the inner walls of the tubular structures.

Each imaging device produces noise. Furthermore, all such systems suffer from nonideal spatial resolution. To avoid artifacts leading to uncertainties in detecting physiological events, a sophisticated filtering procedure is needed. This procedure

should restore the "true" original physiological environment without producing further artifacts. This ability is crucial when processing a sequence of medical images (image segmentation). The post-processing filter used is a robust Wiener filter, as it is applied in many digital image processing devices.² With this procedure, MR Imaging (MRI) and CT images are modified to produce a much better input for further processing and analysis. These filtering algorithms were used successfully for preprocessing MRI sequences in order to reconstruct 3D objects using the Marching Cube³ or Volume Rendering algorithm. The 3D objects were employed for interactive 3D visualization and VIE applications. A scheme illustrating the data flow of the MRI postprocessing is shown in Fig 1.

METHODS AND MATERIALS

MR Data Acquisition for Colonography

The continued high incidence of colon cancer illustrates that the lengthy screening window of 18 to 48 months during which simple polypectomy may be considered curative is not taken advantage of as much as it could be.⁴ In part this reflects the high costs associated with conventional colonoscopy and the poor diagnostic performance of other tests. The search for a screening test permitting a cost-efficient means for detection of polyps has therefore been continued.

MR contrast is based on differences in T1 relaxation times.⁵ Accordingly, for MR-colonography the colon is filled with a water enema spiked with a T1-reducing contrast agent.⁶ As in other colon examinations, MR colonography requires standard oral colonoscopy preparation before the data acquisition. Furthermore, 3D data sets are generally acquired both in the prone and supine patient positions to compensate for residual air in MR data sets. Finally, to reduce bowel motion, scopolamine (20mg) is administered intravenously just before commencing the MR data acquisition.

For MR colonography, a single-contrast enema is administered per rectum using 120 cm of hydrostatic pressure. The

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From the Physics Institute, University of Zurich; MultiMedia Laboratory, Department of Computer Science, University of Zurich; and Department of Diagnostic Radiology, University Hospital Zurich, Switzerland.

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Address reprint requests to Dr Erich Stoll, Physics Institute, University of Zurich-Irchel, Winterthurerstr. 190, CH-8057 Zurich, Switzerland.

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Fig 1. Schematic work schedule of the data postprocessing.

filling of the colon is monitored with a 2D sequence, collecting one image every second. The monitoring sequence also permits the assessment of colonic distention. The administration of scopolamine can be titrated to assure maximal colonic relaxation and distention. Once the enema has reached the cecum and the colon is homogeneously distended, a 3D gradient echo data set, consisting of 60 contiguous coronal 2 mm sections is collected within a single breath lasting under 30 seconds. The technique is based on the use of short repetition (TR 3.8 – 4.7 ms) and echo (TE = 1.6 – 2.0 ms) times rendering most tissues, including fat, darkly. Signal is evident only within the contrast-containing colon. In conjunction with a 40 × 40 cm field-of-view and a 256 × 192 imaging matrix, spatial resolution is almost isotropically defined by a voxel size of 1.7 × 2.0 × 2.0 mm.

The use of a contrast agent in MR colonography adds some cost to the exam. MR does, however, allow monitoring of the colonic filling process, assuring optimal conditions prior to the 3D data acquisition. The latter aids considerably in the subsequent postprocessing of the data sets. Furthermore it allows for the collection of thinner slices in the more efficient coronal plane. Finally, and probably most important, MRI does not require ionizing radiation. The radiation exposure associated with high-resolution CT is considerable, and compounded by the fact that two data sets, one in the prone and a second in the supine position need to be acquired. These aspects have convinced us that the future will belong to MR colonography rather than CT colonography. As long as both techniques remain under investigation, a final answer can not, however, be provided.

Modified Wiener Filtering

The data preprocessing of MRI slices can be improved considerably by using suitable digital filters. Our experiments have shown that about the first and last 5 images of MRI sequences expose a lower level of quality, containing substantially more noise than the rest. Our filtering algorithm is intended to provide the same level of quality throughout all images of a specific MRI sequence.

The filtering procedure is applied on each 2 dimensional slice

supplied by MRI or CT technique. For the sake of simplicity, it is now assumed that the resolution point spread function (PSF) $h(x, y)$ and the noise function $n(x, y)$ are independent from the “true” image function $f(x, y)$. This leads to the linear convolution relation

$$g(x, y) = \int dx' dy' f(x - x', y - y') h(x', y') + n(x, y), \quad (1)$$

which is conveniently transformed into Fourier space,

$$G(\vec{q}) = F(\vec{q})H(\vec{q}) + N(\vec{q}), \quad (2)$$

where capitals denote the Fourier-transformed functions. In principle, this equation can formally be solved for $F(\vec{q})$,

$$F(\vec{q}) = G(\vec{q})/H(\vec{q}) - N(\vec{q})/H(\vec{q}), \quad (3)$$

but the second term should contain full information not only about the amplitude but also about the phase of the noise function $N(\vec{q})$. Since the latter information is not available, the second term in eq. (3) cannot be directly estimated. Furthermore, the noncompensated noise in $G(\vec{q})$ is amplified by division by $H(\vec{q})$ at all wavenumbers q where the PSF $H(\vec{q})$ is small. This property holds for all deconvolution procedures, and everyone is warned not to use such procedures without properly treating the noise.

To minimize the deviation between the surface profile and the reconstructed picture, a robust Wiener filter² has been implemented. Similar filters were also used to restore scanning tunneling microscope pictures by one of the authors.^{7,9} More detailed descriptions have been published.^{2,7-10}

Because the noise does not allow the profile to be restored even if the PSF is known precisely, a filter procedure is applied that leads to the minimum error square between the “true” picture $F(\vec{q})$ and the restored estimate $F_e(\vec{q})$:

$$F_e(\vec{q}) = W(\vec{q})G(\vec{q}). \quad (4)$$

$W(\vec{q})$ is the robust filter transfer function. This function can be written² as

$$W(\vec{q}) = \frac{H^*(\vec{q})\Phi_S(\vec{q})}{|H(\vec{q})|^2\Phi_S(\vec{q}) + \Phi_N(\vec{q})}, \quad (5)$$

where the asterisk denotes complex conjugation, and $\Phi_S(\vec{q})$ and $\Phi_N(\vec{q})$ are the power spectra of the signal and noise processes, respectively.

The usage of the power spectra of the signal and noise means that the large Fourier amplitudes of $|G(\vec{q})|$ are considered without degradation, but small Fourier amplitudes containing mostly noise are suppressed. Whereas the power spectrum $\Phi_S(\vec{q})$ of the signal can be estimated in good approximation by $|G(\vec{q})|^2$ the power spectrum $\Phi_N(\vec{q})$ of the noise is much more difficult estimate. In principle, this can be done self consistently starting with some suitable function $\Phi_{N_0}(\vec{q})$. Instead of $F(\vec{q})$, $F_e(\vec{q})$ is introduced in eq. (2). Then $N_1(\vec{q})$ and $\Phi_{N_1}(\vec{q})$ are computed. These steps are repeated until the new values of $\Phi_{N_i}(\vec{q})$ are deviating less than a given threshold from the older $\Phi_{N_{i-1}}(\vec{q})$ for all $\vec{q}$ -vectors. Unfortunately, this procedure is not stable and nearly never converges. Therefore, for “robust” Wiener filtering,² $\Phi_N(\vec{q})$ is given by a known arbitrary function or constant keeping eq. (5) stable. Here, we have modified the usual Wiener filter algorithm^{2,7-10} where $\Phi_N(\vec{q})$ is proportional to

$\Phi_S(\vec{q})$ based on the assumption justified by statistical laws that the noise spectra are Poissonian distributed. The corresponding amplitudes are therefore proportional to the square-roots of the signal amplitudes. We propose as an estimate for $\Phi_N(\vec{q})$:

$$\Phi_N(\vec{q}) = \phi \cdot (\Phi_S(\vec{q})\Phi_0)^{1/2} \quad (6)$$

with the average power spectrum

$$\Phi_0 = \frac{\sum_{\vec{q} \neq 0} \Phi_S(\vec{q})}{N_q - 1} \quad (7)$$

N_q is the sum of all points used in the the Fourier space. The constant ϕ is supplied as a parameter. For the PSF, we assume a Gaussian

$$h(x, y) = \frac{4}{\pi \Delta r^2} \exp - \frac{x^2 + y^2}{\Delta r^2}, \quad (8)$$

with Δr , the blurring radius of the imaging device, supplied as a second parameter.

In Fig 2, the effect of the filtering procedure is shown, after the most optimal selection of a grey value threshold. In Fig 2a it is very difficult to select a threshold that separates the objects of interest from noise. In Fig 2b this may be achieved more easily.

The output of the image filtering process provides much better input for the Marching Cube³ (3D object reconstruction) and Volume Rendering algorithms (see Fig 1), thus improving the visual output of the corresponding VIE applications.

In addition, the use of the filtering algorithms allows one to reduce the total number of polygons of the resulting 3D objects significantly, thus improving graphics performance and interactivity considerably. Figure 3a shows a surface-rendered 3D endoscopic view of an aorta reconstructed without prefiltered images, Fig 3b shows the same view based on filtered MR images. The total number of polygons (triangles) is reduced from 8388 (a) to 4730 (b).

All images used for this work were acquired on a 1.5 T MR scanner (Signa Echo Speed, GE Medical Systems, Milwaukee, WI). For testing purposes, the data postprocessing was performed in offline mode on dedicated workstations.

Because of the high visualization performance needed for our VIE applications, the filtering algorithm was tested on a Silicon Graphics Onyx 2 workstation equipped with two R10000 CPU's

and 512 megabytes of main memory. Using this hardware platform, the time needed for the filtering process is about one second per image (256×256 pixel resolution).

Figure 4 shows the volume-rendered model of the colon based on raw (a), and prefiltered (b) MR images.

Application: Virtual Endoscopic Viewing

VIE greatly facilitates analysis of MR-colonographic data sets. The complexity of colonic morphology complicates the evaluation of the source data to the point that it may not be interpretable at all. Individual bowel loops overlap, particularly in the pelvic region, and are most difficult to differentiate even when using interactive multiplanar reformation. Particularly small polypoid masses protruding into the colonic lumen are easily missed. This is why VIE has assumed a crucial role in the interpretation of 3D MR colonic data sets.

By providing virtual endoscopic views with 3D depth perception, VIE allows the observer's eye to remain trained on the confines of the colon. Figure 5 shows several views of a VIE session. VIE thus vastly accelerates the data interpretation process and analysis becomes considerably more efficient. Even small structures protruding into the colonic lumen are identified with ease. Furthermore, the 3D depth perception inherent to VIE allows the appreciation of the circumferent morphology of haustral folds, thereby enhancing the ability to differentiate haustra from polyps.

The diagnostic performance of virtual colonography regarding polyp detection is directly related to polyp size. Polyps exceeding 10 mm in diameter are identified reliably. This is evidenced by high sensitivity values beyond 90% reported for both CT and MR colonography.^{11,12} These numbers are based on the acquisition and subsequent analysis of both the prone and supine data sets. Their combined analysis is important in order to cover the entire colonic lumen including those parts obscured by air in MR colonography. The VIE performance is limited however with regard to specificity. Residual fecal material may mimic the appearance of polyps. Detailed analysis of multiplanar reformats frequently helps in this regard. In our practice we thus use VIE as a screen to identify suspicious regions which are subsequently analysed in more detail on the multiplanar reformats.

Fig 2. The segmentation process is demonstrated on a MR colonography slice in coronal plane (a) before and (b) after the application of the new filtering algorithm. Filtering the images enables the generation of virtual colonoscopy scenes with fewer artifacts.

Fig 3. Endoscopic view of a surface rendered MR image of an aorta, (a) before and (b) after the filtering procedure.

DISCUSSION

The robust Wiener filter described above preserves the information of the points with the largest amplitudes in Fourier space. These points consist mostly of true signals and only tiny noise. Points with small amplitudes are dominated by noise and are therefore suppressed. This filter is used to restore all image slices. This facilitates the image segmentation process considerably, thus allowing an improved use of the corresponding postprocessing algorithms (eg, 3D object reconstruction and Volume Rendering). Therefore, the visual output of

the VIE application based on these algorithms gains a much better level of quality.

Because this filter algorithm can not annihilate the noise at a specific point in the Fourier space and preserve the desired signal at the same point, the total information of the image will always be reduced. But the transformation of the image information into Fourier space allows to separate most of this information into Fourier space regions containing predominating valuable signals and others containing nearly only misleading noise. The Fourier transformation is very simple to apply and

Fig 4. 3D view of a volume rendered MR image of the colon, (a) before and (b) after the filtering procedure.

Fig 5. Virtual colonoscopy provides an intraluminal perspective of the colon and enables the visualization of haustra.

runs with high performance on today's personal computers or workstations.

The most suitable choice of the filter parameters ϕ and Δr^2 in eqs. (6) and (8) can be estimated through an interactive visual inspection of the filtered images on the computer display. When the parameter values are too small, the picture is hampered with so called "salt and pepper" noise. On the other hand, when anatomical details tend to be blurred and to disappear, the filter parameters have to be reduced. Unfortunately, probability theory based methods that may be integrated for an automatic noise suppression are very difficult to be formulated, and the simple visual inspection is easy to handle and leads to very reliable results.

Virtual endoscopy will offer new diagnostic opportunities in the assessment of the colon and other tubular structured organs. Generally based on cross-sectional data sets acquired with CT or MRI, it is natural for diagnostic radiology and gastroenterology to be at the forefront of its development. Currently, virtual colonoscopy is considered the most promising VR application. The clinical ben-

efits are a priori, though clinical validation will be needed. Early success will be in those organ systems of large size that have no intrinsic motion and which maintain a lumen without special assistance or preparation. Areas that require special attention include the upper GI tract, vascular tree, and especially the colon.

For virtual colonography to become effective and evolve into an accepted clinical tool potentially obviating the need for diagnostic colonoscopy, it is crucial for radiologists and gastroenterologists to be involved in this early development phase. "Endoscopy" expertise is needed to shape postprocessing algorithms and streamline the analysis approach. VIE will therefore only have a future if tackled together by a team of computer scientists, radiologists and gastroenterologists.

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