Accurate Dynamic Susceptibility Contrast MR Perfusion Quantification using Spatiotemporal Noise Filtering Algorithms

J. C. Kosior[1,2], R. Frayne[2,3]

[1]Electrical and Computer Engineering, University of Calgary, Calgary, Alberta, Canada, [2]Seaman Family MR Research Centre, Foothills Medical Centre, Calgary Health Region, Calgary, Alberta, Canada, [3]Radiology and Clinical Neurosciences, University of Calgary, Calgary, Alberta, Canada

Introduction:

Quantitative cerebral blood flow (CBF) measurements obtained from dynamic susceptibility contrast (DSC) MR imaging are subject to variation due to non-stationary noise artifacts in the post-processed contrast-agent concentration versus time data.[1] These errors cause deconvolution algorithms to become unstable, which often necessitates noise filtering either by (1) explicitly filtering the DSC-MR data prior to deconvolution or (2) implicitly filtering within the deconvolution process. The first method often uses linear spatial filters, which can have the adverse effect of removing important spatial information necessary to localize hemodynamic disturbances. On the other hand, deconvolution algorithms only process 1D temporal information to estimate the noise to be filtered and can only window the post-processed concentration data that contains non-stationary noise concentrated near the peak.[1] Murase *et al.*[2] investigated non-linear anisotropic diffusion (AD) filtering applied to DSC-MR data and reported an improvement in the reliability of CBF measurements. Tomasi *et al.* [3] proposed the bilateral filter as an alternative edge-preserving filter that has since been applied to fMRI data.[4] The purpose of this work is to seek a noise-filtering algorithm that reduces the impact of non-stationary noise artifacts in DSC-MR perfusion data and subsequent errors in CBF estimates. Our hypothesis is that spatiotemporal non-linear filtering will improve the accuracy of CBF estimates while maintaining sufficient hemodynamic tissue contrast for ischemic lesion discrimination.

Methods:

We created a novel digital anthropomorphic perfusion phantom to analyze the accuracy of the noise filters on CBF quantification that consisted of an anthropomorphic brain volume in which each voxel had a known temporal DSC-MR signal based on the tissue content within that voxel. Rician noise was added to the temporal signals to generate a signal-to-noise ratio (SNR) of 20 in white matter (WM). We applied 4D-Gaussian, 4D-AD, and 4D-bilateral filters to the phantom DSC-MR data set using the Insight Segmentation and Registration Toolkit (ITK, National Library of Medicine). The temporal and spatial distance weightings were set to the TR of the sequence and the voxel dimensions, respectively. The Gaussian kernel had a standard deviation (SD) width of 1.88 mm. Tukey's biweight function was used as the AD filter edge-stopping function with the conductance parameter set to $1.48 \cdot \mathrm{MAD}(\nabla \mathrm{I})$ where $\mathrm{MAD}(\nabla \mathrm{I})$ is the median absolute deviation of the gradient magnitude image.[2] Gaussian kernels were used for the domain and range kernels of the bilateral filter. The domain SD width was 1.88 mm and the range SD was set to the SD of the image histogram. We also applied the filters to a patient DSC-MR data set (TR/TE/Flip = 1750 ms/30 ms/45°, 144 × 144 acquisition matrix, 24 cm FOV). PerfTool [5] was used to generate CBF maps using singular value decomposition and an arterial input function (AIF) in the middle cerebral artery. CBF values were rescaled such that normal WM was 22 ml/min/100g to compensate for AIF partial-volume errors. We compared the CBF root mean square error (RMSE) globally for the image and in WM, gray matter (GM), and the known phantom stroke volume (SV) shown in Fig 1a.

Results:

CBF images for the anthropomorphic perfusion phantom are shown in Fig 1 for truth, unfiltered (SNR = 20), 4D-Gaussian, 4D-AD and 4D-bilateral filtering. RMSE values for WM, GM and the SV are annotated within the images. Fig 2 shows the results from filtering the patient data along with annotated CBF values for normal WM and GM and the ischemic lesion.

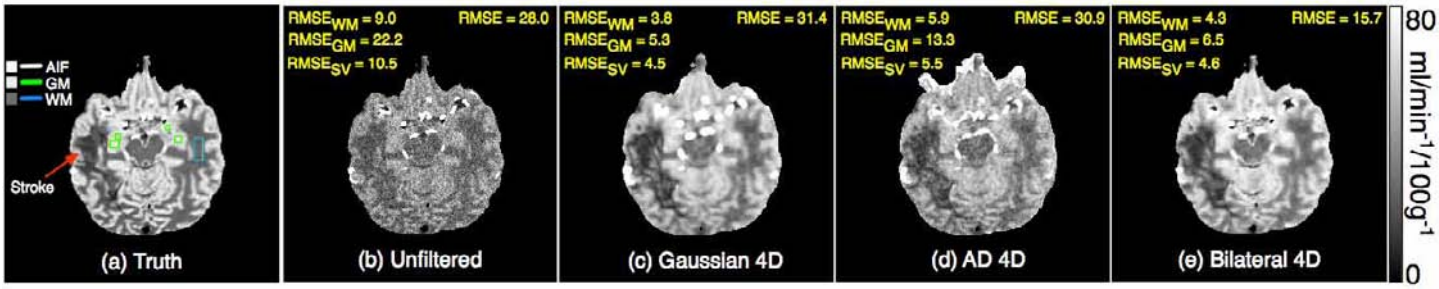

Fig 1: Anthropomorphic phantom. Compared to the truth CBF image (a), the unfiltered CBF (b) has highly variable CBF values and the ischemic region is not visible. The Gaussian (c), AD (d) and the bilateral (e) filtered images more faithfully reproduce the true CBF pattern from a SNR of 20 and the ischemic lesion is clearly visible using all techniques. However, the bilateral filter has overall higher conspicuity near tissue boundaries.

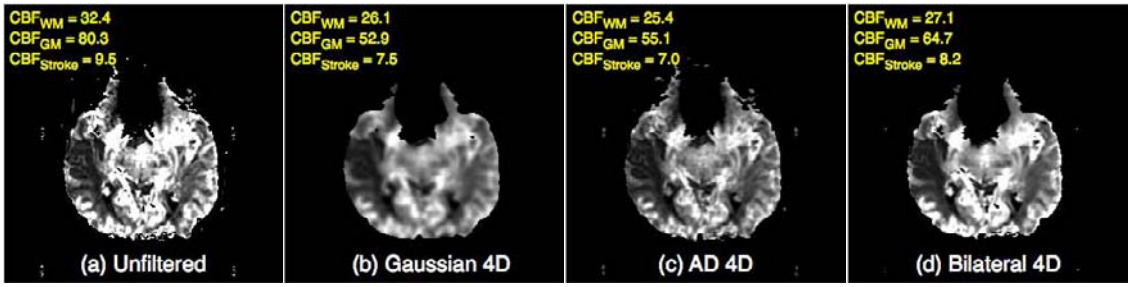

Fig 2: An acute stroke patient imaged using DSC-MR with an ischemic region in the right hemisphere. The filtered images (b-d) differ from the unfiltered image (a) mainly in high-flow tissues.

Discussion

The accuracy of these noise-filtering algorithms was quantified using a novel anthropomorphic perfusion phantom containing known tissue heterogeneity and perfusion. The results of this study demonstrate the importance of removing noise in order to improve CBF estimation, especially in the case of ischemia, as the ischemic region was not discernible in the unfiltered phantom image (Fig 1b). Overall, the Gaussian filter provided the most accurate CBF values for the WM and GM ROIs and the stroke lesion. However, the bilateral filter performed nearly as well but had a significantly lower global RMSE, which is due to better performance within heterogeneous tissue regions where larges fluctuations in CBF occur over small areas (*i.e.*, near arteries) The AD filter did not perform as well as the other filters, and in fact, amplified the arterial signals (Fig 1d) compared to truth (Fig 1a). Robust methods to guide parameter selection for the non-linear filters may improve algorithm performance and are essential before they can become part of routine clinical practice. The simplicity and performance of the Gaussian filter make it an accurate and robust filtering technique.

References: [1] Smith MR *et al.* Mag Reson Med 2003;**49**:122-8 [2] Murase K *et al.* Phys Med Biol, 2001;**46**(10):2713-23 [3] Tomasi C and Manduchi R. Proc IEEE Comp Vision 1998;839-46 [4] Walker SA *et al*. Proc ISMRM 2005;435. [5] Kosior and Frayne. Proc ISMRM 2005;1122.