DENOISING OF PERFUSION ARTERIAL SPIN LABELING DATA BY WAVELET-DOMAIN FILTERING

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<u>Purpose</u>

To decrease the number of averages and the acquisition time required in perfusion arterial spin labeling (ASL) images by wavelet-based noise reduction.

<u>Introduction</u>

Pulsed arterial spin labeling (pASL) MRI is a noninvasive technique for generating quantitative perfusion maps using arterial water as an endogenous tracer. The ASL perfusion maps are based on a subtraction between a tagged image, where the inflowing spins are inverted by short inversion pulses, and a control image. The difference image is directly proportional to blood perfusion. The inversed spins in the tagged image contribute to a very small signal decrease, i.e. the ASL perfusion maps suffer from very low SNR and the experiment must be repeated a number of times (typically around 70) in order to achieve an adequate image quality. In the present study, denoising by a wavelet-based filtering scheme (1) is proposed for reduction of the required number of averages and correspondingly the acquisition time.

<u>Methods</u>

The experimental ASL images were acquired at a 3T MRI unit (Magnetom Allegra, Siemens AG, Erlangen, Germany) using a PICORE Q2TIPS pulse sequence (3) provided by the manufacturer. The imaging parameters were as follows: $TI_1/TI_{1s}/TI_2$=700/1300/1400 ms, TE/TR = 15/2500 ms, FOV=224 mm, slice thickness = 7 mm, 10 cm inversion slab, 1 cm gap and a gradient-echo EPI readout acquisition with 64^2 matrix size. To validate the usefulness of the filtering, a simulated image set was also designed (Fig. 3). In the simulated data, the SNR in one difference map was set to correspond to a realistic ASL case, i.e. approximately 2 in grey matter (GM) and 0.8 in white matter (WM) with nominal CBF values of 25 ml/100g/min in WM and 65 ml/100g/min in GM. The employed wavelet-domain filter has been described previously (1), although presently modified to use a noise-variance assessment according to Alexander et al. (4). Using the notations in Ref. (1), $\mathbf{W_1}$ was the Haar wavelet, $\mathbf{W_2}$ was Daubechies of order 12 and $\mathbf{W_3}$ was Daubechies of order 5. The threshold factor in the hard-threshold filtering part was $\rho=2$. The filter was applied to the averaged difference images (experimental as well as simulated) in which noise can be assumed to be Gaussian distributed (2).

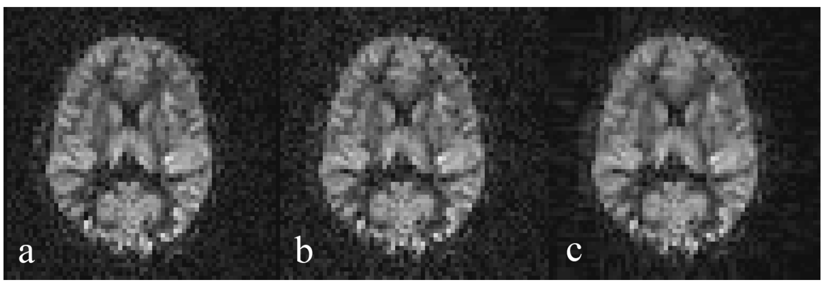

Figure 1. Average image computed from (a) 70 non-filtered difference images, (b) 35 non-filtered difference images, (c) the filtered difference image based on 35 averages.

<u>Results</u>

Figure 1a shows (a) a reference image, based on 70 difference images, (b) the average image computed from 35 difference images and (c) the corresponding filtered image. Fig. 2 shows the calculated CBF values (with error bars) for the simulated data (original and denoised). The results for GM are shown in the left diagram and for WM in the right diagram.

<u>Discussion</u>

The results indicate that by applying the proposed filtering scheme the number of averages and the acquisition time can be reduced by at least 50% with retained standard deviation and with negligible effect on the absolute CBF values. Minor filtering-induced smoothing was observed in the small white-matter structures in the simulated images, represented as high frequencies in the wavelet domain (similarly to the noise). In experimental images, however, this effect is likely to be concealed by normal partial-volume effects. At higher spatial resolution the smoothing artefacts should be minimized.

<u>References</u>

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4. M. E. Alexander et al., MRI 18, 1129-1134, 2000

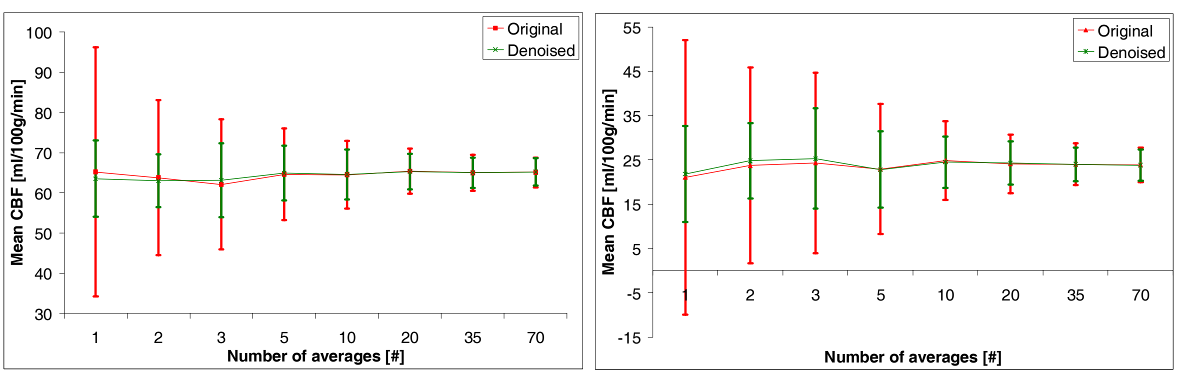

Figure 2. Mean CBF values, with error bars (±1 SD), in GM (left) and in WM (right), for original and denoised images, as a function of the number of averages. Note the decreased SD after denoising, while the absolute CBF values remain unchanged.

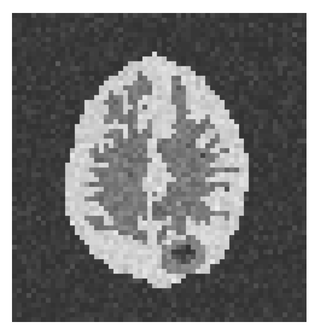

Figure 3. Simulated image example (based on 35 averages).