Reduction of Random and Systematic Errors in ASL Quantitative Perfusion Maps using image Denoising

J. Wells[1,2], D. L. Thomas[1], A. Connelly[3,4], M. King[2], M. F. Lythgoe[2], and F. Calamante[3,4]

[1]Wellcome Trust High Field Research Laboratory, Department of Medical Physics and Bioengineering, University College London, London, United Kingdom, [2]RCS Unit of Biophysics, UCL Institute of Child Health, University College London, London, United Kingdom, [3]Brain Research Institute, Melbourne, Victoria, Australia, [4]Department of Medicine, University of Melbourne, Melbourne, Victoria, Australia

Introduction

Arterial Spin Labelling (ASL) is a totally non invasive technique used in MRI to measure cerebral blood flow (CBF). It is a powerful tool in the understanding and treatment of stroke where studies have suggested ASL CBF maps can provide accurate identification of potentially salvageable brain tissue following an ischemic attack [1]. Despite its current limited use ASL is likely to become more prevalent in clinical and research applications given the continued progression of modern scanners [2]. However, the relatively low cerebral blood volume, and the T1 decay of the endogenous tracer means perfusion maps suffer from poor SNR. Achieving an adequate SNR at a suitable spatial resolution usually requires considerable data averaging. However, in most applications there is a practical limit on the acquisition time. Denoising methods have been exploited in many MRI applications, although their use in ASL has been mainly restricted to fMRI modelling [3]. Given the high SNR demands in ASL, they could form an essential pre-processing step to increase its robustness. In this work we investigate the performance of various common denoising methods in reducing random and systematic errors in ASL CBF and arterial transit time (δa) measures.

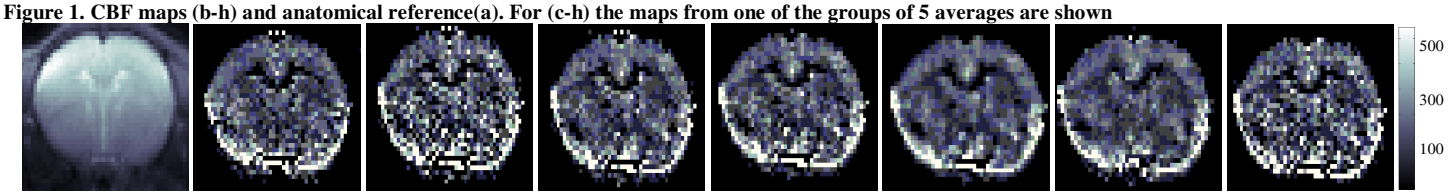
Materials and Method

MRI studies were performed on an anaesthetised male Sprague Dawley (120g) rat using a 2.35T horizontal bore magnet interfaced to a SMIS console, using a volume coil transmitter and a passively decoupled surface coil for signal reception. A continuous arterial spin labelling (CASL) sequence [4] was implemented with a 3 second adiabatic spin tagging pulse applied to a plane 2mm caudal to the cerebellum, perpendicular to the carotid and vertebral arteries. Single slice coronal images, 0.3mm caudal to the bregma, were then acquired using spin echo EPI after a delay time, w. For robust and accurate quantification, it is necessary to measure the perfusion-weighted signal over the time course of the delivery of the tagged bolus. Therefore 10 delay times (w = 0.1, 0.2, 0.3, 0.4, 0.5, 0.6, 0.7, 0.8, 0.9 and 1s) were employed for each tag/control pair. The protocol was repeated for a total of 20 tag/control acquisitions at each w resulting in a total scan time of 25 minutes. Acquisition parameters were: slice thickness = 2mm; image matrix size = 128x64; field of view = 40 x 20mm; TE = 36ms. TR = 2s. The raw data before subtraction were averaged across the entire experiment for each w to produce a high SNR data set; these data were treated as the 'gold standard'. The cortical SNR ranged from 116 to 152 as w increases. The data were divided into 4 groups, each containing 5 of the 20 averages. 'Noisy' images were generated from each of these 4 groups with a typical SNR of 64 to 85. Various common filtering techniques (2D and 2D+t Gaussian filtering, 2D median filtering, 2D Wiener filtering and 2D anisotropic diffusion filtering) were then applied to the 4 noisy data sets. Before and after denoising, the images were pair-wise subtracted to form the ΔM perfusion weighted images. By measuring ΔM over a range of w, CBF maps can be generated by pixel to pixel fitting to the model proposed by Alsop *et* al. [4]. With the current assumptions, analyses of the high SNR data will yield accurate CBF and arterial transit time maps. To quantify differences between the 'gold standard' CBF distribution and those calculated from low SNR data that were denoised, we examined the extent of their agreement by determining the sum of the squared intensity differences (SSD) [5]: $SSD = (1/n)\sqrt{\sum_{p\in\varphi}(I_1(p)-I_2(p))^2}$ where I_1 and I_2 are the two images being compared, and the pixel $\mathbf{p}\in\varphi$ (a large ROI across the cortex containing n pixels under comparison). The SSD tends to zero as the denoised CBF maps approach the 'gold standard' map. Random and systematic errors will be reflected by an elevated SSD; however, it should be noted that some of the random errors may be reduced at the expense of bias. The mean of the ROI in the CBF map provides a suitable indication of possible bias. The CBF(±SE) and SSD are given in ml/100g/min, and the mean results from the 4 groups of 5 averages are reported here.

Results

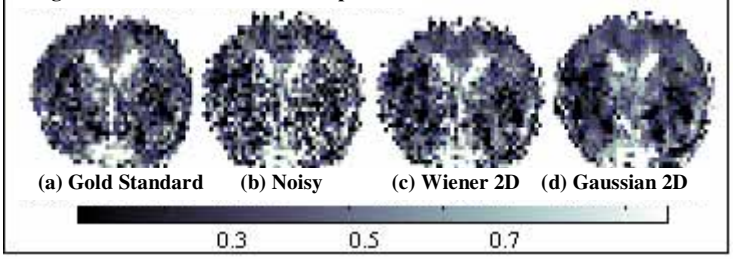
The denoising methods introduced varying degrees of spatial/temporal smoothing to the data. This was reflected in the quality of the calculated CBF maps (see Fig.1). The 2D Wiener (1d), Gaussian 2D (1e) and Anisotropic diffusion (1h) filters appear to increase the agreement to the high SNR map (1b) (cf. results from 'noisy' image (1c)). Application of the Gaussian 2D +t (1f) and Median 2D (1g) filters decreased SSD at the expense of more extreme bias, as shown by the mean values.

Figure 1. CBF maps (b-h) and anatomical reference(a). For (c-h) the maps from one of the groups of 5 averages are shown



(a) Anatomical ref.	(b) Gold standard	(c) Noisy	(d) Wiener 2D	(e) Gaussian 2D	(f) Gaussian 2D+t	(g) Median 2D	(h) Anisotropic 2D
	CBF = 276	SSD= 7.2 CBF = 286±4	SSD = 5.5 CBF = 275±3	SSD = 5.4 CBF =277±4	SSD = 4.9 CBF = 268±4	SSD = 6.0 CBF = 267±5	SSD = 6.7 CBF = 281±3

Figure 2. Arterial transit time maps



(a) Gold Standard (b) Noisy (c) Wiener 2D (d) Gaussian 2D

The denoising methods had also a significant effect on mapping arterial transit time (e.g. see Fig. 2). Visual inspection reveals the Wiener (2c) and Gaussian 2D (2d) filters achieve improved agreement between the noisy arterial transit time maps (2b) to the gold standard δa image (2a).

Discussion

The results of this study demonstrate the potential benefits of pre-processing denoising algorithms in ASL application. As expected some methods are better with regard to structural degradation, minimising random errors, or introducing CBF bias; each of these issues must be taken into consideration in filter selection. For example, the sophisticated spatially adaptive algorithm of the Wiener filter serves to avoid undue bias making it possible to acquire lower SNR data without significant degradation of the CBF maps. In addition, suitable pre-processing may make mapping of arterial transit time more feasible in practical acquisition times. Although in the current study the Wiener and Gaussian 2D filters appear to perform best overall, in general the optimal filter will depend on the particular ASL dataset (e.g. number of averages and number of w) and application (e.g. measuring low CBF vs high CBF), which will determine the degree of spatial/temporal filtering required. It should be noted that the findings of this study should be extensible to human studies as well as pulsed ASL (PASL) techniques.

References: [1] Chalela *et* al. Stroke 2000; 31: 680-7. [2] Petersen *et* al. Br J Radiol. 2006; 79: 688-701. [3] Restom. *et.*al Neuroimage, 2006; 31: 1104-1115. [4] Alsop, Detre. J Cereb Blood Flow Metab 1996; 16:1236-49. [5] Parker *et.*al. J Magn. Reson. Imaging; 2000; 11:702-10.