Adaptive Multi Band Frequency Filter (aMBF) for Noise Reduction in Dynamic CT Perfusion

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Introduction

Dynamic CT perfusion (CTP) consists in a repetitive acquisition of the same body region during a sharp contrast media bolus injection. If motion is corrected for, the profile of CT values over time for each voxel (time-attenuation curve, TAC) can be processed to obtain functional properties, associated with local hemodynamics. These information have demonstrated to play crucial roles in a variety of clinical applications, among which neurology, for stroke management, and oncology, for therapy response monitoring. To limit the otherwise excessive radiation dose, CTP datasets are normally acquired with low kVp and low mAs settings, resulting in a noise level in the same magnitude as the dynamic enhancement to be evaluated.

Materials and methods

Frequency splitting technique has been applied successfully in different ways for CTP datasets. The simplest and most straight-forward implementation of this method is the multi-band frequency splitting filter (MBF[1]), where the high spatial frequencies (HF) are first averaged over time with a fixed width box car function and then combined with the low spatial frequencies of the unfiltered image. In this way, a new dataset is obtained, with reduced noise, but still containing perfusion signal. The main assumption of the MBF approach is that the change over time of the high spatial frequencies is exclusively due to noise. This assumption is almost never fulfilled, since small vessels are also part of high spatial frequencies, and if they would get averaged over time, the arterial input function (AIF) might be compromised. Furthermore, edges are also part of HF images, and if motion is still present, this would be (wrongly) averaged over time. We propose to use the temporal autocorrelation (AC) of the HF images to detect whether a voxel contains mainly noise (AC coefficients with 0 mean) or it contains some sort of non-stochastic pattern, like motion or contrast media dynamics. For each voxel, the absolute value of the sum of the first ten (D=10) AC coefficients is calculated (1) and the temporal Gaussian filter of the HF images is locally adjusted according to this value (2). We named the filter adaptive MBF (aMBF). The scheme is shown in Fig. 1.

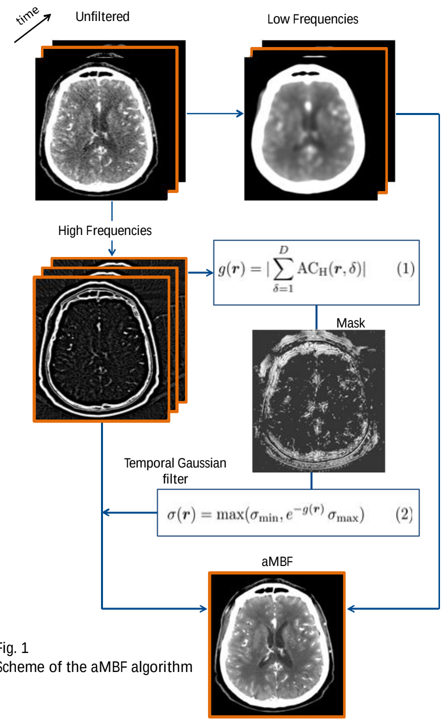

Fig. 1
Scheme of the aMBF algorithm

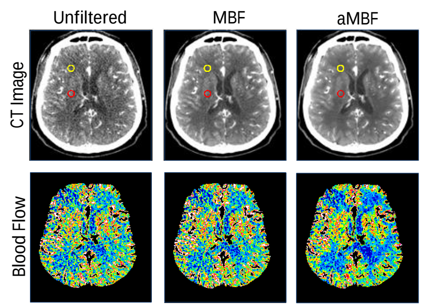

Fig. 2
Example of results obtained via the MBF and the aMBF on a clinical case. ROIs are shown in red for GM and yellow for WM. Blood flow maps show how many mL of blood enter in 100 mL of volume during 1 minute (mL / 100 mL min).

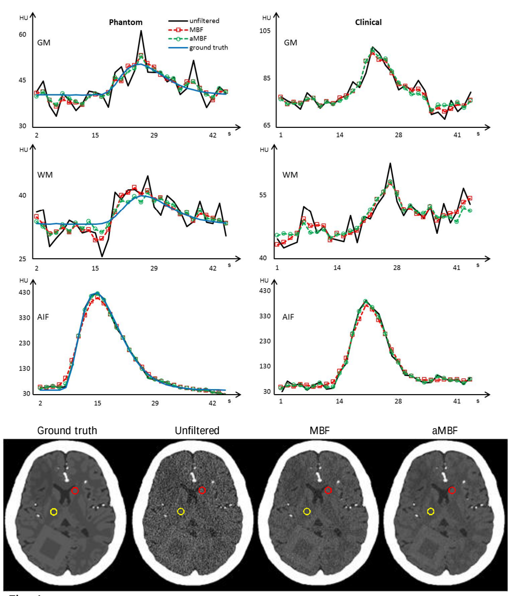

Fig. 4
Top: TACs of digital phantom and clinical case. Bottom: digital phantom results. ROIs are place in gray matter (red) and white matter (yellow).

Results

We tested our method on one digital phantom and on 10 non selected clinical cases, and compared our results with the MBF. In the phantom, the deviation (RMSE) from the ground truth for different types of tissue was systematically lower in the aMBF compared to the MBF, with the major benefits for the vessels (almost 50% error reduction for the AIF). Also in the clinical cases, TACs were smoother but sharper for the aMBF, and CNR was improved on an average by 40% when compared to the MBF.

Conclusions

In conclusion we noted better noise reduction and true signal preservation with the aMBF when compared to the MBF, in similar computational times. Spatial resolution loss was also evaluated and it was similarly negligible in both the two approaches, since no spatial convolution is actually involved.

	Unfiltered	MBF	aMBF
GM	3.95	1.64	1.5
WM	3.2	1.67	1.19
AIF	8	13.33	7.79

Tab. 1
RMSE from the phantom ground truth for different typo of tissue: gray matter (GM), white matter (WM), arterial input function (AIF).

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