# Reduction of Contrast Recirculation Effects in DSC MR CBF Quantification Using Frequency-Domain Modeling

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## Introduction

The accurate assessment of cerebral perfusion is invaluable in the prediction and treatment of cerebrovascular diseases.[1] In dynamic susceptibility contrast (DSC) magnetic resonance (MR) perfusion imaging, cerebral blood flow (CBF) estimates are obtained from the scaled residue function, $R(t)$, determined through the discrete deconvolution of the tissue volume-of-interest concentration, $c_{VOI}(t)$, by the arterial concentration, $c_a(t)$.[1] However, acquisition and post-processing limitations strongly influence the accuracy of CBF estimates obtained from both the widely used time-domain singular-value decomposition (SVD)[1] and the frequency-domain Fourier transform (FT) deconvolution algorithms.[1] In the SVD method, high-frequency noise is suppressed by removing singular values using the threshold parameter $P_{SVD}$. This noise suppression leads to distortion of high-frequency components of $R(t)$, and hence CBF estimation errors, which are largest for short tissue mean transit time (MTT).[1,2] The signal distortions are more easily visualized in the residue spectrum $R(f)$, which is the Fourier transform of $R(t)$. We have previously demonstrated the recovery of distorted residue data by modeling $R(f)$ in the frequency domain, using the frequency-domain Lorentzian modeling (FDLM) approach.[2] We now focus on the performance of frequency-domain modeling, in comparison to the SVD method, for DSC signals containing post-contrast baseline distortion due to contrast recirculation, commonly found in patient data.

## Methods

The FDLM method first involves the selection of relatively undistorted low-frequency residue data, on condition that they are greater than the noise threshold $v$ times the central peak of the discrete residue spectrum ($R'[m\Delta F]$). This data is then linearized and fit to the Lorentzian model. The slope of the resulting linear fit provides the CBF estimate.[2] The tissue concentration function were generated using a mono-exponential residue function.[1] Contrast recirculation was simulated at a delay of 8 s and a dispersion time constant of 30 s.[1] The performance of the FDLM and the SVD method were compared, without and with noise. Monte Carlo noise simulations ($N$ = 1000) were performed for a MR signal intensity signal-to-noise ratio (SNR) of 50, representing high noise. The reformulated SVD method [3] was used, with $P_{SVD}$ = 0.2 for both noiseless and noisy data. The FDLM noise threshold, $v$, was 0.2 for noiseless, and 0.25 for noisy simulations. Patient studies used data obtained at 3 T from three representative ischemic stroke patients.

## Results

Fig. 1a shows an example of baseline distortion caused by recirculation, with the corresponding frequency-domain data shown in Fig. 1b. In the noise-free simulation (Fig. 2a), while a substantially larger CBF underestimation was seen in the SVD results, the accuracy of the FDLM remained closer to the level without recirculation. At MTT > 10 s, increasing CBF estimation error was observed in the FDLM CBF values with recirculation, though the error was less than for SVD. With recirculation, the SVD CBF underestimation can be as high as 50%. At SNR = 50, similar trends were observed. The FDLM CBF estimates displayed larger variations, but remained more accurate than SVD estimates for all evaluated MTT values. The lower sensitivity of FDLM to contrast recirculation was confirmed in patient studies.

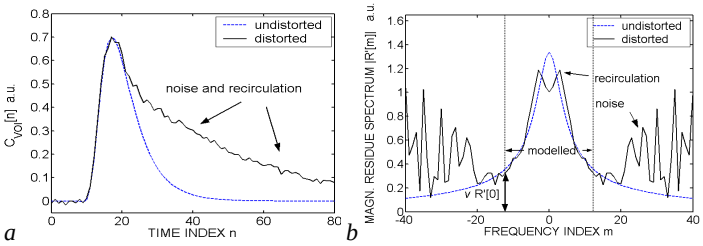


*Figure 1. (a) Distortions due to recirculation can be seen in the low-frequency data of the residue spectrum (b), in this case for MTT =6 s, and can be recovered using the FDLM method. A longer MTT implies a narrower residue spectrum, where the distortion of low-frequency points would have more impact on the accuracy of the model.*

## Conclusion

Contrast recirculation resulted in a noticeable deterioration in the CBF estimation accuracy of the SVD deconvolution algorithm. Frequency-domain modeling is shown to be less susceptible to signal distortions caused by recirculation, as demonstrated through the parametric FDLM method. This may be explained by the observation that recirculation only distorts a small portion of low-frequency data in the residue spectrum, as seen in Fig. 1b. The increasing FDLM CBF error found for long mean transit times may be caused by an increasing low-frequency content of $R(t)$ overlapping with recirculation. The accuracy of the FDLM method could be improved by increasing its robustness to noise through the use of adaptive noise thresholds. This study will also involve further validation on patient data, and lead to improvement of non-parametric frequency-domain modeling.[4]

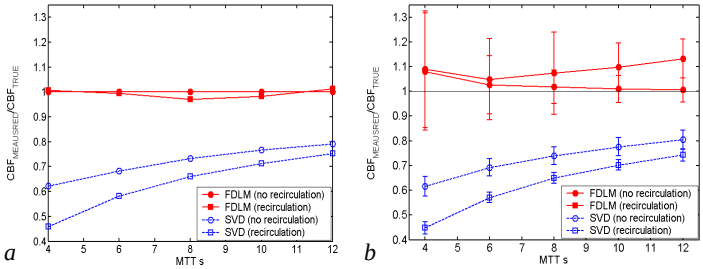


*Figure 2. (a) Noiseless and (b) noisy (SNR=50) simulations show the higher average CBF estimation accuracy of the FDLM method, and its low sensitivity to contrast recirculation, compared to the SVD method, especially at short MTT's. Error bars represent 1 standard deviation.*

## References

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