SHAPE OF THE DSC RESIDUE DATA ACUTE IN ISCHEMIC STROKE

Jeffry R Alger[1], David S Liebeskind[1], Noriko Salamon[2], Jeffrey L. Saver[1], and Danny JJ Wang[1]

[1]Neurology, Geffen School of Medicine at UCLA, Los Angeles, CA, United States, [2]Radiological Sciences, Geffen School of Medicine at UCLA, Los Angeles, CA, United States

Introduction

The standard approach to postprocessing of DSC data is to 1) estimate concentration time data throughout the image volume 2) choose a region from which a global arterial input function (AIF) is defined 3) use singular values decomposition (SVD) to estimate the temporal residue (i.e. the washout that would be observed in the instance of an infinitely sharp arterial input) for each voxel and 4) to estimate various hemodynamic parameters (e.g. CBF) from the residue array's characteristics. SVD is used because it requires no prior knowledge of the shape of the residue data. However use of SVD has the drawback of sensitivity to noise. Filtering by arbitrary truncation of the singular values is used to suppress noise-related oscillation of the derived residue data, but doing so systematically alters the derived hemodynamic parameters values (e.g. CBF) and this necessitates a further secondary correction. Given the limitation of truncated SVD it makes sense to consider alternative means of deriving the residue data. Doing this would be facilitated by prior knowledge of the shape of the residue data. For this reason we undertook a study of the shape of the residue data produced by truncated SVD. The hypothesis is that truncated SVD would provide enough general information about the residue shape to allow the development of model-based deconvolution approaches in the future.

Methods

For this exploratory study we felt it was important to work with very high quality data. We identified DSC studies done using a 3 Tesla MRI unit at our institution for a clinical indication of acute ischemic stroke on day 1 of the current cerebrovascular event. Acquisition parameters were TR (time resolution): 1920 msec, TE: 32 msec, nominal voxel size: $1.7 \times 1.7 \times 6$ mm^3, slices per volume: 26. The studies used a standard contrast agent dose was delivered to an antecubital vein at a rate of 3-5 cm^3 per sec during the dynamic image acquisition. Studies showing poor contrast delivery (AIF half width > 8 sec) were excluded. Studies showing pronounced head movement were also excluded. This process yielded 10 high quality DSC studies.
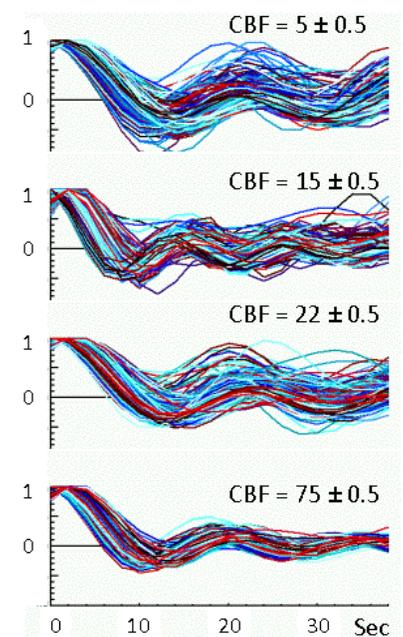

Figure 1: Residue plots obtained from block recirculant deconvolution for multiple voxels in the four characteristic CBF ranges. Each of the four plots comes from a different randomly selected patient. Each voxel's residue data is presented in a different color. All

Data were processed as described above. Each data set was reprocessed with SVD truncation factors that varied from 0.05 to 0.3 using block recirculant and standard SVD. Four-dimensional residue data, and 3-dimensional images of brain mask, CBF, Tmax and CBV were saved for each variation in processing methodology. The typical single subject data set produces approximately 20,000 1-dimensional residue arrays. Because the data set is so large we limited evaluation to voxels having CBF in four characteristic CBF ranges (5 ± 0.5, 15 ± 0.5, 22 ± 0.5, 75 ± 0.5 cm^3 per 100 g per min). These ranges roughly correspond to ischemic core, ischemic penumbra, normal white matter, normal gray matter. We also limited evaluation to voxels that had Tmax (the time of residue maximum) to less than 2 sec. Voxels meeting these criteria were identified in each of the 10 image data sets for each of the processing methods. Residue data were plotted and visually inspected. Mean residue arrays for each subject also were computed and inspected.

Results

In Figure 1 are plotted residue data from multiple voxels having CBF in the four characteristic CBF ranges from four randomly selected subjects. These plots illustrate a generally consistent shape that appears to be a slightly delayed Gaussian washout convolved with a variable oscillatory function. A few voxel residue curves in each set showed pronounced variation in the frequency and the amplitude of the oscillatory component. At very low flow rates the oscillatory component tended to overwhelm the Gaussian washout component. The oscillatory component is likely the result of truncation of the dynamic concentration data for both artery and tissue. Neither dynamic concentration will typically return to baseline by the conclusion of dynamic image acquisition because of systemic steady state labeling of the vascular system. Figure 2 shows voxel averaged residue data for each of the CBF ranges for each subject. The findings for the different SVD truncation levels and deconvolution methods were consistent with those illustrated in the figures, although higher truncation levels produced smoother residue data as expected (data not shown). The figure illustrates that the washout half time for all CBF ranges in all subjects is 5 - 8 sec. ANOVA failed to find a significant relationship between the washout half time and CBF across the subjects.
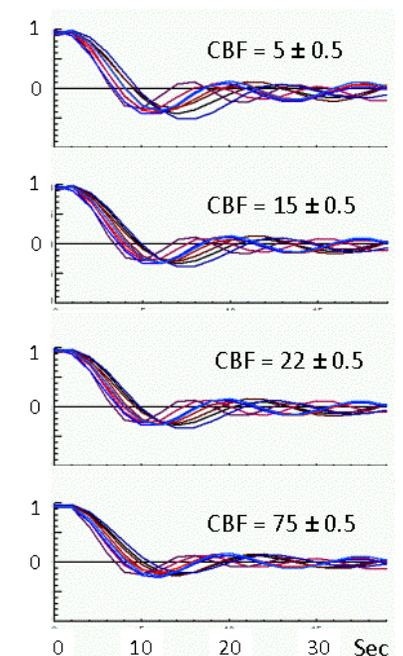

Figure 2: Residue plots of voxel averaged data for all 10 subjects obtained with block recirculant deconvolution. Color is used to denote different subjects' data.

Conclusions

These results suggest the DSC residue data could be adequately modeled as a (possibly time shifted) Gaussian shaped washout function having a half time of 5 – 8 sec and that such a modeling would be appropriate across acute stroke patients over a wide range of CBF values, which include those typically found in ischemic core and penumbra. The absence of a strong relationship between CBF and residue washout time indicates that nature achieves higher per unit mass flow rates by increasing the blood volume rather than by increasing the rate of volume flow through the brain's microvessel system. Overall these findings suggest that appropriately designed model-based deconvolution approaches that include prior knowledge of shape as discussed above may prove to be a feasible alternative to SVD deconvolution in DSC data processing.