Digital Anthropomorphic Perfusion Phantom for the Evaluation of DSC-MR Perfusion Algorithms

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Introduction

Many different post-processing algorithms can be applied to dynamic susceptibility contrast (DSC) MR perfusion data to derive as cerebral blood flow (CBF) maps. It is critical to validate these algorithms, since physicians rely on CBF maps to deduce knowledge about abnormal cerebral hemodynamics for conditions like acute stroke.[1] Patient data alone cannot be used to validate the accuracy of these algorithms, since truth is unknown. Previous studies have relied mainly on simulated 1D signals to evaluate different post-processing strategies.[2,3] However, spatiotemporal algorithms (*i.e.*, noise filters, registration methods *etc*.) require realistic anatomical information in order to validate their impact on CBF estimation.

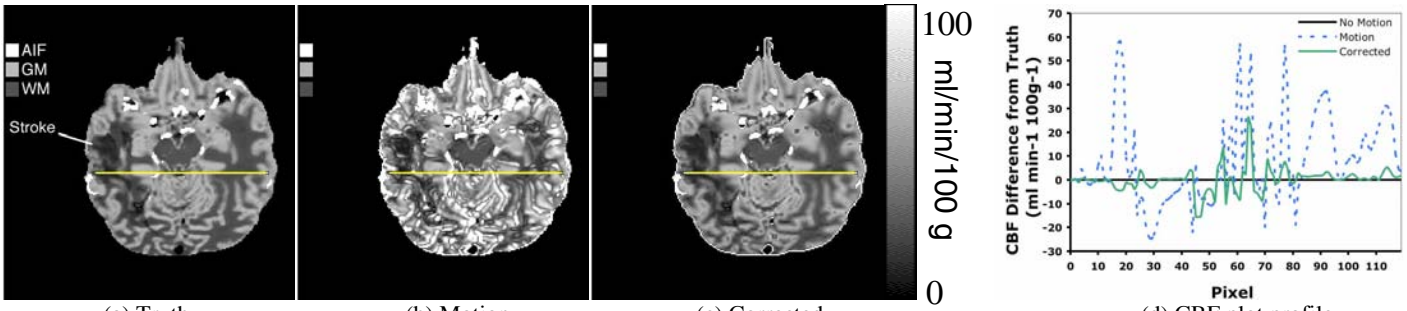
Collins *et al.* constructed a 3D digital brain phantom to assess image-processing algorithms with a particular focus on multiple sclerosis.[4] Although this data set contains realistic tissue volume fractions (TVF), it does not include blood vessels, which are necessary for arterial input functions (AIF). The purpose of this work was to construct a digital anthropomorphic perfusion phantom that can be used to evaluate the accuracy and robustness of CBF estimates obtained using different algorithms in the presence of simulated acquisition artifacts (*i.e.*, noise). This data set was constructed similarly to the temporal DSC-MR perfusion test pattern concept proposed by Kosior *et al.*,[5] except that we also added anthropomorphic spatial information based on real patient data. Hence, the phantom consisted of an anthropomorphic volume where each voxel had a known temporal DSC-MR signal based on the tissue characteristics within the voxel and the acquisition artifacts under investigation.

Methods

We empirically obtained the inversion times (TI) necessary to null the signal from white matter (WM), gray matter (GM) and cerebrospinal fluid (CSF) at 3 T (Signa; GE Healthcare, Waukesha, WI) by sweeping through a range of TIs using a T2-weighted inversion recovery (IR) fast spin-echo (FSE) sequence (TR/TE/Flip = 8000 ms/99 ms/90°, 256 × 128 acquisition matrix, 24 cm FOV). Using these TIs, we collected three IR-FSE data sets plus an acquisition with no inversion (TR/TE/Flip = 7250 ms/14 ms/90°, 144 × 144 acquisition matrix, 24 cm FOV, 2.5 mm slices with no gaps). For each IR-FSE image, we obtained pure measurements of WM, GM and CSF using the mean intensities from regions of interest (ROI) placed in the corpus collosum, the basil ganglia and the ventricles, respectively. These IR-FSE images and the ROI measurements we used to generate four equations for each voxel of the form $S_{voxel} = \alpha_{WM} \cdot S_{WM} + \alpha_{GM} \cdot S_{GM} + \alpha_{CSF} \cdot S_{CSF}$ where S_{tissue} and α_{tissue} are the pure tissue ROI measurements and the tissue volume fractions (TVF), respectively. We generated a fifth equation by noting that the TVFs sum to unity. We solved this over-constrained system of equations in a least-squares sense using singular value decomposition (SVD) to obtain the TVFs for WM, GM and CSF. To acquire the TVF for bulk blood (*i.e.*, arteries), we acquired 2D time-of-flight MR angiography (MRA) images with the same coverage as the IR-FSE acquisitions (TR/TE/Flip = 40 ms/3 ms/30°, other parameters the same as IR-FSE). We thresholded the arterial signals and then normalized them to generate a fuzzy mask representing the TVF for blood. The total static TVFs for each voxel were adjusted using the blood TVF such that all four tissues summed to unity. Using the TVF data set, we then generated simulated perfusion signals for each voxel using previous simulation approaches (*i.e.*, gamma-variate arterial flow, mono-exponential residue function).[2,3]. Rician noise was simulated by adding white noise in quadrature (*i.e.*, assuming the T2* signal was purely real) to generate different SNRs relative to WM. A stroke region was added by (1) co-registering CBF images from a stroke patient to the phantom, (2) thresholding the patient CBF images to isolate an ischemic lesion, (3) normalizing the lesion to create a fuzzy mask representing the degree of ischemia, and (4) simulating ischemic signals using the fuzzy mask. To demonstrate the utility of this phantom, we analyzed the effect that patient motion has on CBF using a noise-free phantom. A few sequential temporal volumes were rotated axially (total range < 5°) during the bolus passage to simulate typical patient motion artifacts. The individual 3D temporal volumes of the DSC-MR data set were aligned using rigid-body registration (SPM). CBF maps for no-motion, motion and motion-corrected images were generated using PerfTool.[5]

Results

The TIs for WM, GM and CSF were found to be 550 ms, 900 ms and 2250 ms, respectively. CBF images generated using the anthropomorphic perfusion phantom for no motion, motion and motion-corrected are shown in the Figure along with a CBF profile for the horizontal line in each image. The root mean square errors of the plot profile for motion and motion-corrected were 20.5 ml/min/100 g and 5.4 ml/min/100g, respectively.



(a) Truth (b) Motion (c) Corrected (d) CBF plot profile

Figure: Example of patient motion artifacts. Compared to the truth CBF image (a) there are large errors in CBF where temporal signals from different tissues overlap near edges in the case of axial motion. Realignment (c) shows that CBF estimates improve. CBF slice profiles (horizontal line in (a)-(c)) reveal that motion errors are most severe near tissue boundaries and that temporal realignment drastically reduces these errors.

Discussion

The anthropomorphic perfusion phantom enabled us to analyze the effect of patient motion on CBF along with the subsequent temporal volume realignment correction using registration methods comparable to that used in some vendor software. Scientists can process the anthropomorphic perfusion test pattern interactively like regular patient data to rapidly visualize and quantify (since truth is known) the performance of new and existing vendor algorithms in the presence of different artifacts (*i.e.*, noise, AIF partial-volume errors, temporal-aliasing *etc*.).

References

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