Assessing cerebrospinal fluid flow in the brain using 3D phase contrast velocity encoded MRI

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Introduction. Phase contrast (PC) velocity encoded MRI has widely been used to analyse cerebrospinal fluid (CSF) flow in the brain [1-3]. Up until now PC flow measurements of CSF in the brain have only been done in two dimensions (2D). By using a three dimensional (3D) PC pulse sequence, flow in the entire CSF system can be assessed in one scan and errors associated with imaging CSF pathways in oblique planes can be eliminated. The PC pulse sequence was evaluated for slow flow using a flow phantom and used to assess CSF flow *in-vivo* on normal volunteers. *In-vivo* communicating flow between ventricles and CSF passageways are identified using rules of nearest-neighbour connectivity within a 3D integrated flow volume on five healthy volunteers.

Methods. All studies were performed using a 3T MR-scanner (Magnetom Allegra, Siemens Medical Solutions, Erlangen, Germany) and a single channel head coil. An RF-spoiled gradient echo 3D sequence with interleaved 3-directional velocity encoding was used for the *in-vivo* studies and the sequence was adjusted for 2D imaging with 1-directional through-plane velocity encoding for the phantom validation [4]. The steady state flow phantom consisted of a bottle containing stationary water encircled by a soft silicone tube containing carefully controlled flowing water. Through-plane flow velocities between 5.8 and 76.0 mm/s were measured with the following imaging parameters: TE = 4.4-10.9 ms; TR = 7.1-13.6 ms; flip angle = 15 °; spatial resolution = 0.9×0.9 mm^2; slice thickness = 3.2 mm; matrix size = 256×256. The VENC was varied from 0.01 to 0.20 m/s, and cardiac gating was not applied since the flow was constant, resulting in scan times between 13.9 and 7.3 s. For lower VENCs phase aliasing occurred and the region of interest (ROI) was phase unwrapped using Goldstein's 2D branch cut algorithm [5]. The velocity of the flowing water was calculated using $v = (\Delta\phi / \pi)VENC$ where v is the velocity and $\Delta\phi$ is the phase from the phase difference image.

For the *in-vivo* studies, five normal male adults were scanned using the following image parameters: TE = 6.4 ms; TR = 9.2 ms; VENC = 0.04 m/s; flip angle = 15 °; spatial resolution = 1.6×1.6×1.6 mm^3; temporal resolution = 73.6 ms; matrix size = 128×128; slices = 20-24. The scan time was around 14 minutes. The imaged volume was centred on a midline sagittal position and covered the entire width of the 4th ventricle and the majority of the lateral ventricles. The measurements were prospectively gated to the ECG and the number of cardiac phases varied between 10 and 12. As a result of high gradient strength associated with the low VENC, induced eddy currents and magnetic field nonlinearities resulted in phase aliasing in the phase and frequency encoding directions. A ROI was manually drawn on a magnitude image and this region for the first time point in each slice and each velocity encoding direction was spatially unwrapped in 2D using Goldstein's branch cut algorithm. These unwrapped images were then used to unwrap the rest of the data in one dimension through time. One dimensional temporal unwrapping is made possible since the motion of the brain and CSF cavities is small and is applied because of its speed and simplicity. A second order polynomial surface was then fitted to the unwrapped images and then subtracted from the unwrapped images to remove background phase errors due to field inhomogeneities. To extract and highlight only voxels with significant flow information, the data from the three velocity encoding directions was combined. If X_n, Y_n, and Z_n correspond to the 3D volumes for the three encoding directions at time point n, an integrated flow volume can be created using Equation 1 where N is the total number of time points. A threshold was selected with reference to the noise level of stationary tissue in V. A region in the brainstem was selected and a threshold of two standard deviations above the mean noise was selected. It should be noted that both flow direction and temporal information are lost when calculating V, but these can be ignored for the sake of purely assessing flow connectivity. The connectivity of flow throughout the CSF system was calculated by clustering pixels in V that are connected in 26 adjacent voxel neighbourhoods. The connectivity results make it easy to automatically identify any occlusions along the CSF pathways. The flow data was visualized by computing isosurfaces from V by connecting points with pixel intensities above the threshold value. All post-processing was written in Matlab (The MathWorks Inc., Natick, MA).

$$V = \sum_{n=1}^{N} \sqrt{X_n^2 + Y_n^2 + Z_n^2}$$

Equation 1

Results and Conclusions. In all 5 volunteers 26 nearest-neighbour connectivity of CSF flow from the lateral ventricles to foramen of Magendie could be shown. A typical computed 3D isosurface is shown in Figure 1A and the corresponding anatomical reference is shown in Figure 1B. It should be emphasized that the volume image is based entirely on flow data. This method allows for a 3D semi-automated non-invasive assessment of the CSF pathways inside the brain. Future work will aim at assessing flow outside the brain in the subarachnoid spaces and extracting quantitative flow data from 3D images.

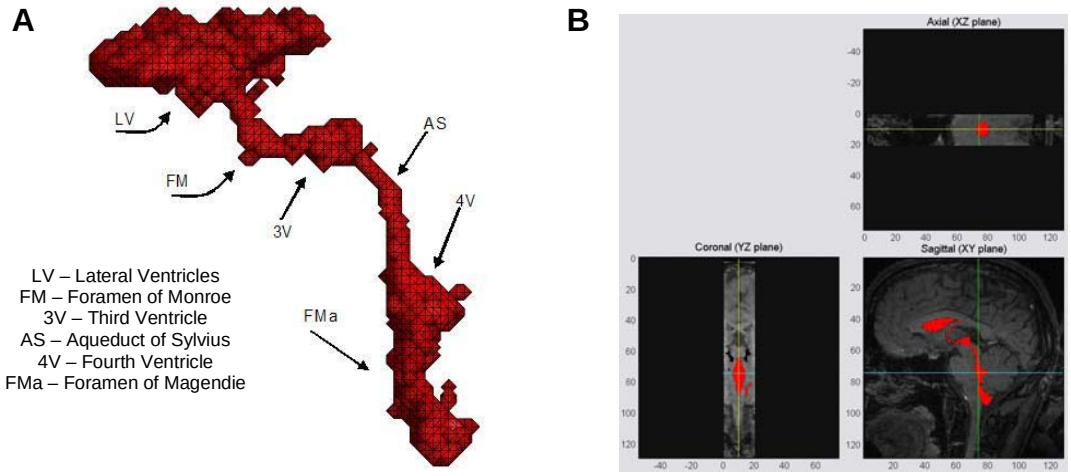

Figure 1. A) *3D volume image of the CSF system inside the brain showing connectivity of flow from the lateral ventricles to the foramen of Magendie.* B) *Anatomical reference for image in* A. *Sagittal midline slice and Axial and Coronal slices through 4V are shown.*

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4. Markl *et al.* JMRI 2007;25:824–831. 5. Ghiglia and Pritt. Two-dimensional phase unwrapping: theory, algorithms and software, Wiley-Interscience, 1998.