Multi-point velocity encoding for simultaneous assessment of arterial, venous and cerebrospinal flow

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Introduction: The acquisition of arterial and venous blood flow to the cranium as well as flow of cerebrospinal fluid (CSF) in the spinal canal are the main input parameters for modeling cerebrospinal dynamics [1]. Phase-Contrast (PC) MRI has been used to measure input and output boundary conditions to populate model parameters. There are, however, limitations as to the accuracy of PC-MRI when mapping arterial influx and venous outflow with conventional phase-contrast methods. While arterial blood enters the scull in feet-head direction through the common carotid (CCA) and vertebral arteries (VA), the venous system is geometrically more complex without a well-defined principal axis of flow and with large inter-subject variation [2]. The jugular (JV) and vertebral veins are accompanied by collateral pathways that can be numerous, small in diameter and along various orientations. Arterial inflow and venous outflow is often measured using the method introduced by Alperin [3]. According to this method, arterial blood flow and flow in the jugular vein are measured over one cardiac cycle with two PC-MRI acquisitions with high and low velocity encoding. As net arterial blood flow has to equal net venous blood flow, venous blood flow is scaled up to match arterial inflow. The simultaneous acquisition in a dual v_{enc} approach has been proposed in order to detect velocity information at the same time points [4]. Despite dual v_{enc} acquisition, velocity sensitivity might still be insufficient when attempting flow measurements in the venous system beyond the jugular veins.
The present study aimed at simultaneous measurement of arterial, venous and CSF flow in the neck using a multi-point variable-density velocity encoded 3D sequence with spatiotemporal undersampling.

Methods: A 3D volume in the neck of 3 healthy volunteers was measured using a gradient-echo sequence on a 3T Philips Achieva System (Philips Healthcare, Best, The Netherlands). Flow encoding gradients were applied in feet-head direction corresponding to v_{enc} = [128, 64, 32, 16, 8] cm/s. A FOV of 128x128x10mm^3 was sampled isotropically with 0.8mm resolution and a temporal resolution of 30-49ms. With 8-fold k-t undersampling and [10x6] training profiles in [k_yxk_z], scan time was 5:30 to 8:42min depending on heart rate. Images were reconstructed using k-t PCA [5]. Additionally, a 2D PC slice was acquired fully sampled with $v_{enc,high}$ = 128cm/s and $v_{enc,low}$ = 16cm/s in the location of the central slice of the 3D volume with 0.8x0.8mm^2 in plane resolution and 5mm slice thickness. Scan time in this case was 2:10 to 3:48min. Velocities v were assessed by Bayesian parameter estimation [6] according to the signal model $S(k_v) = S_0 \cdot e^{-\frac{\sigma^2 k_v^2}{2}} \cdot e^{-ivk_v}$ [7].
Arteries and veins in the neck were segmented following the scheme in Fig. 1. Static tissue was identified using Otsu's segmentation method [8] on the magnitude images. Vessels were segmented based on the complex signal difference between velocity encoding segments. The segmentation results were further adjusted based on the stroke volume (SV) per voxel computed as the flow rate integrated over time. Median filtering was applied to suppress remaining salt and pepper noise. The threshold level was set at 2x standard deviation of the static tissue signal.

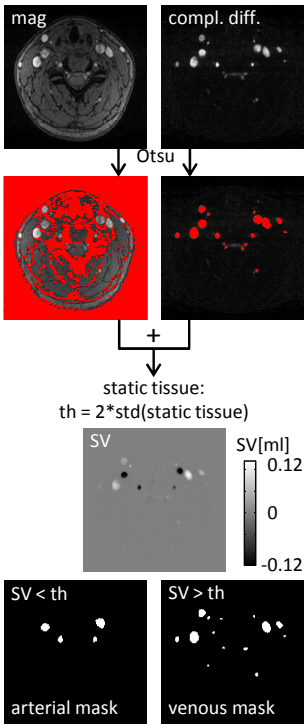

Fig 1: Segmentation of arteries and veins from one slice of the 3D-multi-k_v velocity field.

Results: Per subject data is given in Tab.1. The intra-scan comparison of flow differences in arterial and venous blood flow shows an improvement for the 3D multi-k_v acquisition. Differences in SV between arteries and veins were 0.9 - 11.4% and 10.6 - 22.5% for the 3D-multi-k_v and 2D PC acquisitions, respectively. The inter-scan comparison of arterial SV acquired with multi-k_v 3D method and the 2D PC reference shows differences of 0.6-2.4ml. The absolute CSF SVs were compared to the $v_{enc,low}$ PC acquisition. The difference in SV of CSF as results of small velocities is 0.04-0.3 ml. Fig. 2 show individual drainage pattern with vessel wise pulsatility. The pulsatile shape is not identical in all vessels.

Tab. 1: Stroke Volume in different vessels and flow difference

		Stroke Volume [ml]				tot. no. of veins	flow diff. [%]	abs. SV CSF [ml]
		arterial	left JV	right JV	tot. venous			
V1	PC-2D	11.82	2.74	5.13	10.57	8	10.54	1.39
	multi-k_v3D	13.71	3.8	5.83	13.83	14	0.85	1.35
V2	PC-2D	11.3	0.35	3.93	8.76	14	22.48	0.73
	multi-k_v3D	13.73	1.36	4.55	12.17	17	11.41	0.41
V3	PC-2D	23.12	2.55	0.18	19.12	13	17.32	1.92
	multi-k_v3D	23.76	1.81	0.81	22.39	27	5.76	1.88

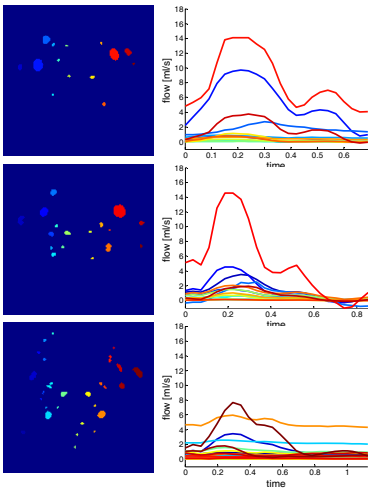

Fig 2: Individual venous drainage pattern for 3 different volunteers with different venous pulsatility in different vessels. The first volunteer has prominent jugular vein flow while the third volunteer shows only little drainage through the jugular veins.

Discussion: The feasibility of simultaneously measuring arterial, venous and CSF flow in the neck using a 3D multi-point velocity encoding scheme has been demonstrated. Due to inter-subject variability of venous drainage pattern, the prediction of the optimal velocity sensitivity range is difficult, and accordingly sequences covering a large range of velocities are preferred. As it is not possible to place a 2D measurement plane orthogonal to all veins, the use of 3D thin slice acquisition is essential to map the normal velocity component for SV determination. The combined velocity field for high and low values resulting from Bayesian velocity estimation facilitates segmentation of all arteries and those veins that are within the range of spatial resolution.

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Acknowledgement
SNF K-32K1_120531/1