# Acclerated Dual Venc Phase Contrast VIPR in Healthy Volunteers

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**Introduction:** Rapid imaging techniques now allow for the acquisition of volumetric, time-resolved, phase contrast (PC) images with three directional velocity encoding. Covering large vascular territories with a single 4D PC MRI makes choosing the proper velocity encoding (Venc) challenging. If the Venc is too low for a vessel of interest, velocity aliasing occurs. If chosen too high, the velocity-to-noise ratio (VNR) suffers as VNR ~ 1/Venc. 5-pt velocity encoding has been proposed to address this issue, but the gains in the dynamic Venc range are modest [1].

Previous work demonstrated the benefits of accelerated dual Venc PC VIPR in phantom studies [5] when larger gains in dynamic range are desired, i.e. when imaging flow jets or when measuring velocity fields for peak flow and wall shear stress calculations. It was the aim of this in-vivo study to compare flow measurements derived from different dual venc settings to those obtained with (1) a 4-point balanced and a (2) 5-point velocity encoded PC VIPR acquisition as well as (3) a targeted standard Cartesian 2D phase contrast (2D PC) acquisition.

**Background:** Dual Venc acquisitions can improve the VNR while maintaining dynamic range utilizing (1) a low Venc data set to provide a good VNR and (2) a high Venc data set to identify and correct for velocity aliasing [2, 3]. However, in 4D velocity mapping, this approach significantly lengthens the already long scan duration. For undersampled acquisitions, such as PC VIPR [4], accelerated dual Venc images can be obtained utilizing additional undersampling to capture the high Venc acquisition. Since the high Venc image is only utilized for unwrapping the low Venc small errors from additional undersampling will not influence the final image.

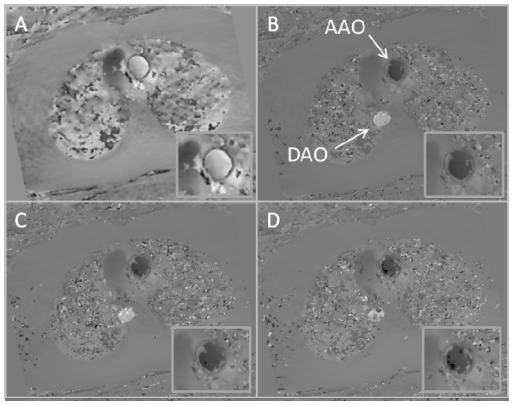


*Figure 1: PC VIPR image of aortic blood flow with Venc = 45cm/s. (A) Uncorrected; (B-D): Corrected images with (B) 100% (C) 50% and (D) 25% of the acquired high Venc projections. Error in unwrapping increases with greater high Venc undersampling (shorter total scan time).*

**Materials and Methods:** *Volunteer protocol:* Ten healthy volunteers (8M/2F, mean age = 27 years) were scanned on a clinical 3T system (Discovery 750, GE Healthcare, Waukesha, WI). Three PC VIPR data sets with balanced velocity encoding gradients [5] were acquired at three Vencs (160cm/s, 80cm/s, and 40cm/s) with the prescription volume covering the entire chest. In addition, a 5-point velocity encoded PC VIPR data set was acquired with a Venc=160cm/s. 2D PC measurements were performed in three locations: perpendicular to the ascending aorta (AAO) at the height of the pulmonary artery (Venc = 160cm/s), main pulmonary artery (MPA) (Venc=120cm/s), and superior (SVC) (Venc=80cm/s). 2D PC images in a static phantom were acquired for background phase correction for residual phase errors [5].
*Acquisition Parameters:* PC VIPR data were acquired with a dual-echo PC VIPR trajectory with respiratory and retrospective cardiac gating. Typical scan parameters were: 1.25 $mm^3$ isotropic spatial resolution, BW = 125 kHz, TR 6.2-6.6ms, volume: 32cm x 32 cm x 20 cm, 15,000 dual echoes, scan time: ~12-13 min (4-pt) and 15min (5-pt). 2D PC data were acquired with a product Cartesian sequence with through-plane velocity encoding, and prospective ECG gating in a breath hold. Typical scan parameters were: 35x35$cm^2$ FOV, 6 mm slice, TR 5ms, matrix = 256x160, BW 62.5kHz, 20 views per segment, scan time = 22 heart beats.
*Reconstruction:* A dual Venc image reconstruction algorithm was implemented that combines data from the low Venc (80 cm/s or 45 cm/s) and the high Venc (160 cm/s) acquisitions to correct phase wraps in the low Venc image. The high Venc data was reconstructed with 100%, 50%, and 25% of the acquired projections to mimic acquisitions of reduced scan times from a single data set. These datasets were combined with the low Venc data with the goal of generating velocity aliasing free data. 2D PC and PC VIPR results were compared by conducting flow measurements in equivalent planes. The PC VIPR flow volumes in each vessel were compared with the 2D PC MR flow volumes for statistical significant differences using the paired t-test ($p<0.05$).

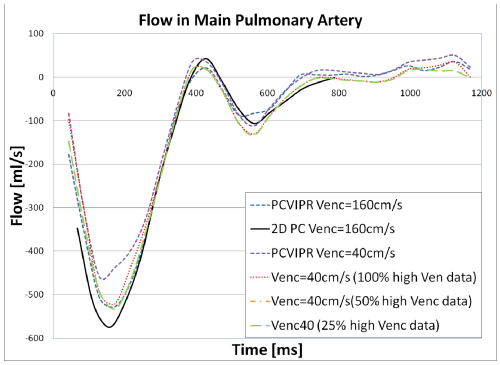


*Figure 2: Flow waveforms measured in the MPA of a healthy volunteer. The low Venc (Venc = 40cm/s) data contains phase wrapping errors which can be corrected with 100%, 50%, or 25% of the high Venc data (Venc = 160cm/s).*

**Results:** Fig. 1 shows a resliced image of aortic blood flow from a PC VIPR acquisition with a Venc=45cm/s. In the uncorrected image, aliasing occurs over the entire aorta. The other images show the same slice after phase correction with 100%, 50%, and 25% of the high Venc (160 cm/s) projections. Flow waveforms from a volunteer measured in the MPA are show in Fig. 2. Errors in the waveform measured from the low Venc PC VIPR image are corrected after phase unwrapping of individual voxel phases.
Table 1 summarizes results of flow and velocity measurements averaged over all 10 volunteers. . Dual Venc processing with a Venc = 40cm/s unwrapped with 25% of the high provided poor results in the AAO because of unresolved phase wrapping errors. All other techniques were in good agreement with 2D PC MR flow in the MPA and SVC and in reasonable agreement in the AAO. Statistically significant differences in the flow volumes were found for: AAO flow volume measurements from PCVIPR Venc=160cm/s, Venc=40cm/s unwrapped with 50% and 25% of high Venc data and Venc=80cm/s unwrapped with 25% high Venc data .

| | | | | Venc=40cm/s | | | Venc= 80cm/s | | |
|---|---|---|---|---|---|---|---|---|---|
| | 2D PC | PCVIPR Venc160 | PCVIPR 5pt Venc160 | 100% | 50% | 25% | 100% | 50% | 25% |
| Relative scan time | | 1 | 1.2 | 2.06 | 1.56 | 1.31 | 2.03 | 1.53 | 1.28 |
| Ascending Aorta | | | | | | | | | |
| Flow Volume (ml) | 87.8 | 75.1 | 72.5 | 68.9 | 67.8 | 55.4 | 88.2 | 84.8 | 84.7 |
| Peak Flow (ml/s) | 403.3 | 376.4 | 353.2 | 371.4 | 371.2 | 237.6 | 406.4 | 387.6 | 368.1 |
| Mean Velocity (cm/s) | 18.4 | 16.3 | 16.1 | 14.8 | 14.6 | 7.2 | 11.4 | 12.4 | 10.7 |
| Main Pulmonary Artery | | | | | | | | | |
| Flow Volume (ml) | 97.4 | 90.7 | 81.5 | 90.7 | 91.9 | 89.5 | 87.0 | 88.4 | 84.8 |
| Peak Flow (ml/s) | 425.4 | 400.2 | 363.1 | 394.1 | 394.8 | 390.7 | 398.4 | 404.3 | 427.9 |
| Mean Velocity (cm/s) | 15.6 | 13.6 | 13.2 | 13.9 | 14.0 | 14.6 | 13.4 | 13.4 | 13.5 |
| Superior Vena Cava | | | | | | | | | |
| Flow Volume (ml) | 31.7 | 30.6 | 29.1 | 31.7 | 31.3 | 31.8 | 30.8 | 31.0 | 32.9 |
| Mean Velocity (cm/s) | 15.5 | 14.6 | 16.6 | 14.4 | 14.5 | 15.0 | 13.6 | 13.3 | 12.8 |

*Table 1: Flow and velocity parameters measured in 10 healthy volunteers. The values show good agreement across the techniques used. In the SVC, the peak flow is not analyzed because in most cases, the venous flow waveforms did not have a distinct peak.*

**Conclusions:** This study demonstrated the feasibility of in-vivo dual Venc PC VIPR acquisitions. With this technique, it is possible to maintain the VNR of a low VENC acquisition while gaining a fourfold increase in velocity range with a 50% increase in scan time. Alternatively, the velocity range can be doubled with only a 25% increase in scan time over balanced-encoded PC VIPR. This technique could be especially useful in cases of disease where stenotic jets can cause very high velocities or where very low velocities are of concern, e.g. in wall shear stress measurements. In future work we want to investigate the flow underestimation in the AAo, which is possibly caused by the temporal filtering used in the reconstruction of the PC VIPR data.

**References**: [1] Johnson *et al. MRM* 63(2), 2010[2]Lee AT, Pike GB, Pelc NJ. *MRM*, 33(1),1995 [3]Johnson KM *et al. Proc ISMRM*, 2006 [4]Gu TL *et al. AJNR* 26(4), 2005 [4]Nett EJ *et al. Proc ISMRM,* 2010 [5] Pelc JMRI 1991;1(4): [6] Chernobelsky *et al. JCMR* 9(4), 2007.