# Cardiac Output Measurement with Ungated Spiral Phase-Contrast and Triggered Real-Time SSFP Imaging

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**Introduction:** Cardiac output (CO) measurement is clinically important in the assessment of patients with cardiovascular diseases. However, current non-invasive in-vivo CO measurements have proven to be elusive because of the need for cardiac-synchronization and relatively long scan-times. In this study, we propose an ungated spiral phase-contrast (USPC) imaging [1] as a fast non-cardiac-synchronized method for measuring CO. USPC is shown to quickly measure accurate volume flow rates within few seconds even in the presence of strong pulsatility and has been successfully applied to renal arteries and peripheral vessels [1,2]. Flow phantom experiments were performed to optimize imaging parameters of USPC, and CO of three normal volunteers were measured with USPC and compared to a stroke volume (SV) calculation with triggered real-time SSFP (RT-SSFP) imaging [3]. The results show good agreement between the two methods.

**Methods:** The USPC pulse sequence uses spiral readouts with pseudo-randomized interleave ordering [4], and a minimum-first-moment flow-encoding (FE) scheme [5] in order to minimize pulsatility artifacts (Fig. 1) [1]. A Kaiser-bessel window shaped RF-pulse of 480us was used to minimize the second-order moment of FE lobes, enabling a minimum TE of 2.2ms with slice thickness of 3mm and Vmax (maximum velocity encoded) of 250cm/s. Correction of eddy-current induced phase-offset at the ascending aorta (AA) by extrapolating phase-offsets from static chest materials is prone to produce an error, and therefore a short TE of 3 ms was used. In addition, FE lobes were played strictly along Z (axial slice) since at our site, the Z-gradient produced the least eddy-current phase-offset. Other scan parameters of USPC were: 12 interleave spirals, TR=12ms, FOV=24cm, $\Delta x$=2mm, and flip(in vivo)=10deg. Various flip angles were tested and exact Vmax was calibrated with a flow pump that produced highly pulsatile flow as in a normal ascending aorta (Pump-1421, Harvard Apparatus, MA). For both pump and in vivo data, spatial interpolation of an image is done to minimize intra- and inter-observer error from regions-of-interest (ROI) segmentation. Although only a few cardiac cycles are necessary to produce an accurate time-averaged flow rate [1], the scan was continuously played out for a 15-second breath-hold to reach a steady-state that minimizes the partial-averaging error of USPC [1]. For in vivo comparison, cardiac-gated movie loops of 10 slices were acquired with RT-SSFP in breath-holds of 10 R-R intervals (Fig. 2). SV was calculated from the difference between end-diastolic and end-systolic volumes, and CO was finally acquired by multiplying SV with the heart-rate.

**Results:** As shown in Fig. 3, a small flip angle can minimize the over-estimation caused by inflow effects (magnitude enhancement of high velocity signals) in ungated PC imaging [6], where the complex summation of temporal velocity distribution in a voxel happens before the quantification. A flip angle of 10 degree still produced sufficient SNR with ungated imaging. In vivo CO results in Fig. 4 show good agreement between USPC and RT-SSFP measurements. A cumulative-average plot of CO in Fig. 5 (acquired from the same raw-data sets) reaches a steady-state, which represents the time-averaged CO. In all subjects, we were able to localize axial slices perpendicular to AAs with an interactive real-time imaging system [7].

**Discussion:** Vessel movement did not cause severe errors in CO measurements (free-breathing results in Fig 4). Theoretically, a vessel compliance or movement will cause partial-volume effects (PVEs) at the edge voxels; these PVEs are caused by the combination of temporal and spatial velocity distribution in the voxels during the scan time. Shorter scan time is possible by using the periodic information in the cumulative-average plot as in Fig. 5. However, to minimize the partial-averaging error associated with this process, at least 3 to 4 cardiac cycles are needed in AA, which can be shown by simulations. In our experiments, the scan time of 15 seconds was used to minimize the partial-averaging error and to speed-up the post-processing time. Vmax twice as large as the peak velocity can minimize the under-estimation caused by the asymmetry error from temporal velocity distribution (i.e., temporal partial-voluming) [1]. However, high Vmax can cause phase-noise errors. Since the scan time of USPC can be only a few seconds, we can perform multiple measurements with different Vmaxs and flip angles to separately estimate the under- and over-estimation caused by temporal partial-voluming and inflow effects, respectively.

**Conclusion:** We have demonstrated a new approach to in vivo CO measurements based on USPC. The results indicate that this method has the potential to quickly and robustly measure CO without cardiac-gating and possibly without breath-holding. In combination with SV measurements, USPC can also be used to rapidly quantify regurgitant volume in patients.

**References:** [1] Park JB, et al., MRM, 49:322-328, 2003. [2] Park JB, et al., Proceedings of ISMRM 2002, p.1800. [3] Nayak KS, et al. MRM, 49:188-192, 2003. [4] Spielman DM, et al., MRM, 34:388-394, 1995. [5] Pelc NJ, et al., MRQ, 10:125-147, 1994. [6] Maier SE, et al., MRM, 34:706-712, 1995. [7] Kerr AB, et al., MRM, 38:355-367, 1997.

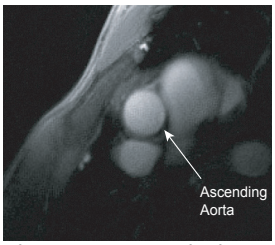


*Fig. 1: USPC magnitude image of AA. Motion- or flow-artifact is minimized.*

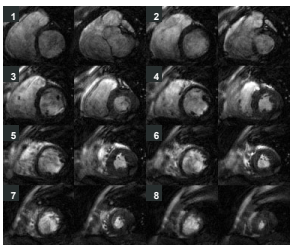


*Fig. 2: RT-SSFP images for SV calculation. End-diastolic (left) and end-systolic (right) frames are shown for the first 8 slices.*

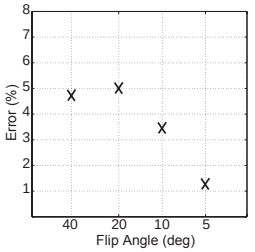


*Fig. 3: Flow pump measurement with USPC. With highly pulsatile flow, small flip angle is necessary to prevent overestimation.*

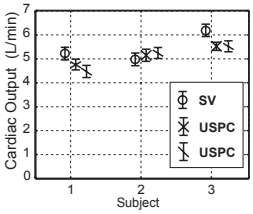


*Fig. 4: CO (L/min) measurements.*

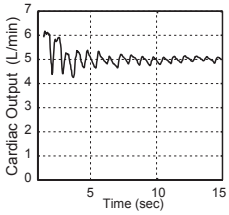


*Fig. 5: USPC cumulative-average plot of CO of subject 2 in Fig. 5, approaching the time-average CO in Fig. 5.*