Multi-Slice CINE Phase Contrast Pulse Wave Velocity Measurements for Characterising Aortic Cardiovascular Disease

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Introduction and Aims: Aortic stiffening occurs as a result of the pathophysiological contribution of age and atherosclerosis, and is associated with increased risk of cardiovascular events. This can be characterised by measurement of pulse wave velocity (PWV). The gold standard PWV is obtained via direct intra-aortic pressure measurement, but these experiments are invasive and procedurally complex. Cardiac MRI (CMR) on the other hand is non-invasive and can interrogate multiple vessel regions within a single examination. This is important in the case of the aorta which differs both structurally and functionally along its substantial length. CMR measurement of PWV has recently been proposed as an imaging marker of arterial stiffness, and can be quantified using cardiac-gated CINE phase contrast techniques acquired at two aortic locations [1, 2]. The aim of this study was to extend this method to derive multi-slice phase contrast MRA data (with 'through plane' velocity encoding) to measure flow and pulse wave velocity calculated over 6-8 different anatomical sites from the aortic arch to the renal bifurcation. The specific objectives were two-fold, namely (i) to investigate the repeatability of using this multi-slice technique to measure PWV in a cohort of healthy volunteers, and (ii) to extend the measurement to derive PWV data from two carefully defined clinical cohorts of patients with 'high risk' or 'known' cardiovascular disease.

Methods: Three clinical cohorts were studied as follows: (i) Young Healthy Normal volunteers (YHN) - 5M, 5F with mean age 27±5 years and no prior history of cardiovascular disease, (ii) Clinical High Risk volunteers (CHR) with elevated plasma brain natriuretic peptide (BNP) – 5M, 5F with mean age 52±7 years, and (iii) patients with known Peripheral Arterial Disease (PAD) – 4M, 3F with mean age 65±11 years. ECG-gated segmented CINE breath-hold phase contrast MRA (PC-MRA) was performed on a 3T Magnetom Trio MRI scanner (Siemens, Erlangen, Germany) to determine PWV. A combination of a body matrix coil and spine array coils was used. Multiple images of the aorta were acquired in axial oblique orientation from the aortic arch (ascending aorta at the level of the pulmonary bifurcation – *figure 1*) and down the descending aorta (until immediately proximal to the renal arteries) using a consistent gap of 3cm between acquisitions. Imaging parameters were TR/TE = 22.85/1.83ms, flip angle 30°, slice thickness 6mm, field of view 350mm, 3 segments, 1 signal average, and in-plane pixel matrix 144x192. Through-plane velocity encoding was prescribed with a VENC of 150cm/sec, and 128 (interpolated) phases per cardiac cycle. The typical acquisition time was approximately 20secs, although dependent upon patient heart rate. The YHN cohort was scanned on two successive occasions over the course of a day in order to derive 'scan-to-scan' repeatability PWV measurements and the two clinical cohorts were scanned at a single time-point.

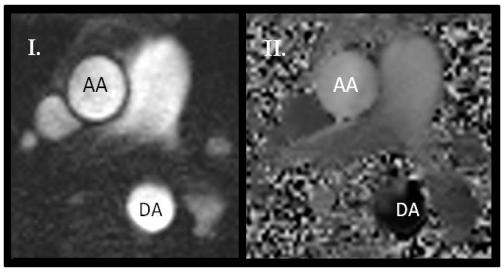

Figure 1. (I) Magnitude and (II) Phase image of the ascending(AA) and descending aorta (DA) taken at the aortic arch

Image processing was performed using *Segment v1.9 R1917* (Heiberg, Germany). Aortic region-of-interest contours were defined at each temporal phase of all datasets resulting in flow waveforms corresponding to each location. The transit time was determined by modeling a linear regression from the flow up-slope and the arrival time for each pulse wave was measured relative to the first (baseline) ascending aorta waveform. The distance of each measurement plane (also relative to the baseline ascending aorta location) was calculated using calipers on an ECG-gated CINE segmented fast low-angle shot (FLASH) dataset acquired in the sagittal oblique 'candy cane' orientation – *figure 2*. A plot of distance (x) versus transit-time (y) for each anatomical location was then able to elicit the PWV, calculated as the inverse gradient of the resulting 'best-fit' line. An example set of flow waveforms acquired across the cardiac cycle (n=8 anatomical positions) for a healthy volunteer is illustrated in *figure 3*.

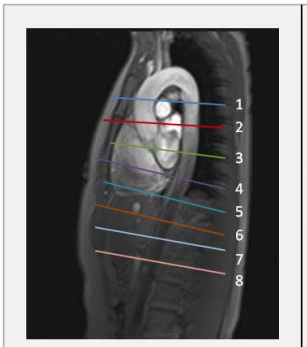

Figure 2. Oblique 'candy cane' orientation of the aorta with slice positions (1-8)

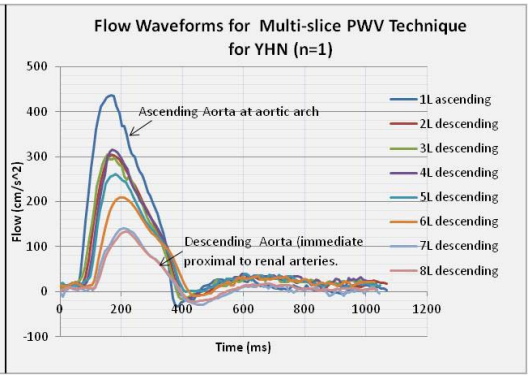

Figure 3. An example set of flow waveforms acquired across the cardiac cycle (n=8) anatomical positions for a healthy volunteer

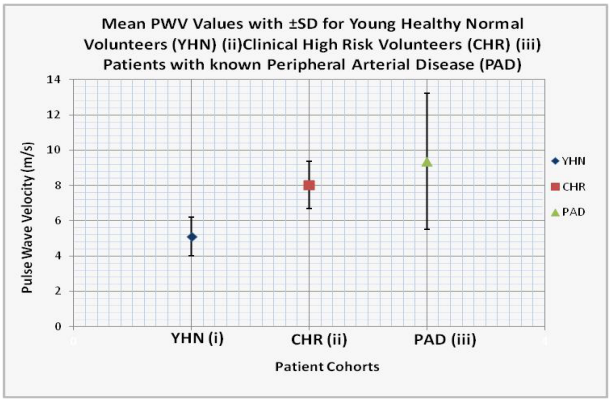

Figure 4: An example set of flow waveforms acquired across the cardiac cycle (n=8) anatomical positions for a healthy volunteer

Results and Discussion: The calculated mean MRI PWV for the YHN cohort (5.2 ± 1.1m/s) was significantly slower relative to the mean PWV for the CHR cohort (8.0±1.4m/s, $p<0.05$) and the PAD cohort (9.4±4.1m/s, $p<0.05$) – *figure* 4. However there was no significant difference between the mean MRI PWV values between the two clinical groups. The spread of PWV values (i.e. larger standard deviation) observed in the PAD group may be explained by the known greater disease heterogeneity (atheroma burden) within this cohort. The mean PWV for the YHN cohort was found to be consistent when measured on three separate occasions (by the same observer) over two imaging sessions, i.e. no significant differences to the mean PWV were identified – even on a 'scan-to-scan' basis.

Conclusion: PWV can be derived from phase contrast MRA using multiple aortic flow waveforms in combination with intra-arterial distance measurements. In this study, the mean PWV was found to be stable on a 'scan-to-scan' basis in a cohort of healthy volunteers, and was found to increase significantly with age and cardiovascular disease severity – i.e. the data are consistent with developing aortic stiffness.

References: [1] Westenberg JM *et al,* JMRI 2011, 13(1), pp.15-17. [2] Rogers, W.J. *et al,* JACC 2001, 38(4), pp.1123-9