Impact of Eddy Currents Correction in Aortic Pulse Wave Velocity Assessment on 4D Flow Magnetic Resonance Images

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***Abstract*— Arterial stiffness is an independent predictor of cardiovascular disease, usually estimated using Pulse Wave Velocity (PWV) measurements. PWV can be calculated estimating the distance between different arterial beds and the time-shift of their respective pressure or flow. Non-invasively, PWV can be calculated by tonometry or with Magnetic Resonance Images (MRI) techniques (2D+t or 3D+t). In 2D+t studies only two measurement sites are possible, whereas in 3D+t, several arbitrary locations can be analyzed. When acquiring velocities in MRI, there is an offset error that needs to be corrected ("zero correction"). The aim of this work was to study the impact of this correction in PWV estimations. We acquired 4D Flow MRI sequences in seven asymptomatic healthy volunteers. We estimated the flow in five sites of the thoracic aorta along with the distance amongst them. The slope of a linear regression model between time-shift and distance allowed the estimation of the PWV. We compared the values before and after the zero correction. Mean and standard deviation of PWV were 11.09 ± 1.80 m/s and 11.88 ± 2.89 m/s, respectively (p=0.23). We concluded that PWV can be assessed in 4D Flow MRI sequences and the zero correction did not modify the PWV values.**

***Keywords*— Pulse Wave Velocity, Aorta, Magnetic Resonance Images, 4D Flow.**

***Resumen*— La rigidez arterial es un predictor independiente de las enfermedades vasculares y es usualmente estimado mediante la Velocidad de Onda de Pulso (VOP). La VOP puede ser calculada estimando la distancia entre distintos segmentos arteriales y el desfasaje temporal de sus respectivas ondas de presión o flujo. No invasivamente, la VOP puede ser calculada utilizando tonometría o mediante técnicas de Imágenes de Resonancia Magnética (IRM) (2D+t o 3D+t). En estudios 2D+t sólo es posible estimar el flujo en dos sitios, mientras que en 3D+t, se pueden analizar diferentes sitios en forma arbitraria. Cuando se adquieren velocidades en IRM, se produce un error de offset que debe ser corregido ("corrección del cero"). El objetivo de este trabajo fue estudiar el impacto de esta corrección en estimaciones de VOP. Adquirimos secuencias de IRM-Flujo 4D en siete voluntarios sanos asintomáticos. Luego, estimamos el flujo en cinco sitios de la aorta torácica, a la vez que la distancia entre ellos. Se realizó una regresión lineal entre el desfasaje de los flujos y la distancia entre los sitios, cuya pendiente permitió estimar la VOP. Comparamos los valores antes y después de la corrección: El promedio y desvío estándar de la VOP fue 11.09 ± 1.80 m/s y 11.88 ± 2.89 m/s, respectivamente (p=0.23). Concluimos que la VOP puede ser estimada en secuencias de IRM-Flujo 4D y que la corrección del cero no modificó los valores de VOP.**

***Palabras clave*— Velocidad de Onda de Pulso, Aorta, Imágenes de Resonancia Magnética, Flujo 4D.**

I. INTRODUCTION

Cardiovascular disease is the major cause of morbidity and mortality in the world. Most cardiovascular pathologies begin with atherosclerosis: development of plaques formed by lipids, fibrous tissue and (eventually) calcium within the arterial wall. At the beginning of this disease, the plaques grow outwards, i.e. without altering arterial lumen [1]. For this reason, early detection of atherosclerosis is not possible with traditional angiographic methods [2]. However, arterial walls become stiffer in the early stages of the disease. Aging has a strong influence on aortic morphology and structure that is accelerated with hypertension [3]. Also, arterial stiffness has been shown to be an independent factor to predict cardiovascular events [4, 5]. Throughout the years, many methods were developed to measure arterial stiffness [6]. In this work we focused on Pulse Wave Velocity (PWV), that is the standard parameter to assess arterial stiffness [7].

PWV is the speed with which the pressure pulse generated at the heart travels through the arterial walls. It

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is usually measured as the distance between two arterial sites divided by the Transit Time (TT) of the corresponding waveforms. The goal standard to measure PWV is via a catheter. Since this procedure is invasive, it is not used in prevention. PWV can be non-invasively estimated by means of Carotid-Femoral PWV (cfPWV). This technique consists in measuring pressure signals by tonometry in the carotid and femoral arteries. Computing the TT and measuring the distance between the sites cfPWV is obtained.

There are some limitations regarding cfPWV that are worth to be mentioned. i) The assessment of pressure by tonometry can be difficult in obese subjects and dangerous in patients with plaques [8]. ii) The measurement of distance between the carotid and femoral arteries is approximated, with high error and low repeatability [9]. iii) The PWV obtained by this method is not a central measurement and correspond to a global stiffness (it represents an average of all arterial beds between the carotid and the femoral).

PWV can be assessed with Magnetic Resonance Images (MRI) studies. Some authors have estimated PWV in a single plane containing the ascending and descending aorta (2D+t) [10]. By means of the Phase Contrast (PC) technique, orthogonal velocities at each point of the vessel lumen can be assessed. Recently, a new sequence called 4D Flow has been introduced (3D+t) [11, 12]. This technique allows the visualization of a volume through time, with the velocities in each point and each instant. In particular, this MRI sequence allows a simultaneous assessment of PWV at different aortic segments. When performing an MRI PC study, phenomena called “zero error” occurs due to the eddy currents. It adds a systematic offset to the velocity estimation that needs to be corrected [13]. The correction of this “zero error” is still controversial in 4D Flow sequences and less studied in PWV estimations.

In this work, we proposed to quantify the impact of the zero correction in 4D Flow MRI studies on thoracic aorta PWV assessments.

II. Materials and Methods

A. Data acquisition

Seven healthy asymptomatic volunteers (aged 54 ± 23 years old, 4 male) underwent an MRI examination with gadolinium contrast agent. We acquired 4D Flow sequences with a sampling rate of 50 Hz and a velocity encoding of 250 cm/s. All subjects rested in supine position for 15 minutes before the acquisition. Diastolic and systolic pressure were 74 ± 8 mmHg and 123 ± 15 mmHg, respectively, and the heart rate was 64 ± 4 beats per minute. The images were obtained using a General Electric 3T MRI scanner. Scanning time was approximately 11 minutes per subject. The acquired volumes were (in average) 129x241x376 mm, digitalized into 127x103x256 voxels. Studies were exported in a DICOM format to be processed later on a computer.

B. Flow measurement

In order to estimate the flow in the aorta, we developed a custom software in C# in our lab. The platform allowed the user to reconstruct any arbitrary oblique plane to delineate a vessel perimeter creating Regions of Interest (ROIs). In these planes, the user was able to visualize the blood velocities and to calculate a cross-sectional area and flow.

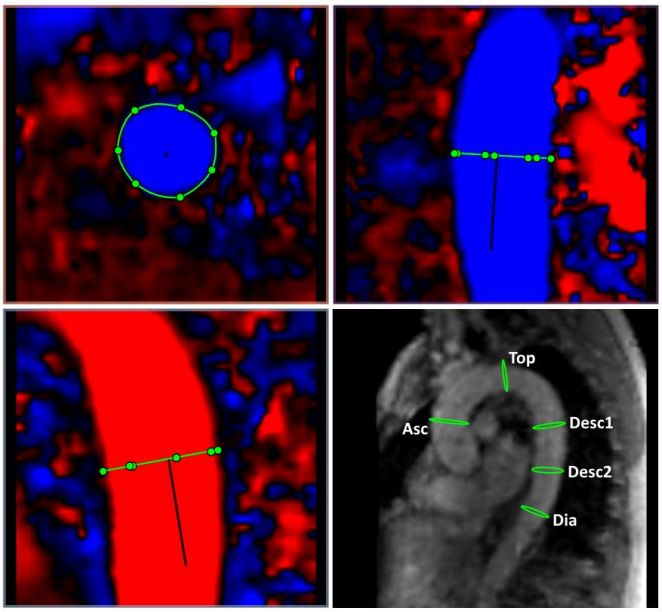

Fig. 1. Bottom left and top: the three orthogonal views of a ROI with the velocities in blue-red scale. In black is the velocity vector at the center of the aorta, that was used to position the plane of observation. Bottom right: an oblique sagittal view, showing the five ROIs selected.

We estimated the flow waveforms at five locations of the aorta: Ascending (Asc), Isthmus (Top), proximal descending (Desc1), distal descending (Desc2) and diaphragm (Dia) (Fig. 1). Asc and Desc1 were measured at the level of the pulmonary artery bifurcation. The Top was placed midway of the previous two. Dia was located at the point where the aorta goes through the diaphragm. Finally, Desc2 was defined as the midpoint between Desc1 and Dia.

For each of the five locations the user manually defined a ROI as follows. First the systole was identified as the instant when the highest velocity was registered in the center of the aorta (observed orthogonally). Then a plane normal to this velocity was selected. Finally, the user manually contoured the aorta by setting approximately 7 to 8 points (Fig. 1). The pinpoints were interpolated with a curve using a closed centripetal Catmull-Rom spline [14]. Eventually, if the aorta moved within the cardiac cycle, several curves were drawn at different times and then they were automatically interpolated for all phases. This occurred specially in the Asc location. In the cases when the aorta remained static through the cardiac cycle (e.g. in the descending aorta), the same curve was used for all times.

We computed the flow $Q(t)$ through each ROI at each time as the sum of the orthogonal velocities in the ROI, times the area of the corresponding pixels (1).

$$Q(t_i) = \sum_{\vec{X} \in ROI(t_i)} \left[\vec{v}(t_i, \vec{X}) \cdot \vec{n}(t_i)\right] \Delta A \quad (1)$$

where t_i represents each time, $\vec{X}$ are the pixels inside the ROI, $\vec{v}(t_i, \vec{X})$ is the velocity at a single time and a single pixel, $\vec{n}(t_i)$ is the unitary normal to the plane of the ROI at the time t_i and ΔA is the area of the pixels. Using (1), we obtained five flow waveforms, one for each location of the aorta.

C. PWV assessment

To calculate PWV we computed the distance between the centers of the ROIs and the TT between the flow waveforms. We estimated the distance between each pair of consecutive ROIs by manually placing points in the centerline of the aorta. The first and last points were placed at the center of each ROI. To estimate the TT, we developed the following algorithm. First, we interpolated the signals with a cubic spline in order to have a 1 kHz reconstructed sampling rate. We detected the maximum (systolic) and the minimum (diastolic) flows, thus defining a min-max range. We determined the upslope of the flow as the portion of the curve between the 20% and the 80% of the min-max range [10, 15]. In this interval we fit a line by minimum squares. The temporal intersection between this line and a zero flow was considered as the flow curve time-shift. The TTs were calculated as the difference between the time-shifts of consecutive waveforms.

With these estimations we obtained the distance between the ROIs as a function of the time-shifts of the corresponding waveforms. PWV was estimated as the slope of the regression model between TT and distances.

The correction of the studies was performed as follows. First, a static region was determined for each sagittal plane. A given voxel was considered static if the three components of the velocity had a standard deviation below the 35th percentile of its plane. Then, a surface was fit to the static points of each plane in order to interpolate the offset in the non-static regions. We calculated the PWV in the corrected and uncorrected studies using the same set of ROIs for each volunteer.

Statistical analysis

In the PWV estimation, the coefficient of correlation was calculated for each slope. To compare PWV values between uncorrected and corrected studies, a paired Student's t-test was used. Values of p lower than 0.05 were considered significant.

III. Results

A representative regression model between TT and distance is shown in Fig. 2. PWV values for uncorrected and corrected studies are shown in Table I. PWV resulted 11.09 ± 1.80 m/s and 11.88 ± 2.89 m/s, respectively. Non-significant differences were found between uncorrected and corrected studies (p=0.23). The correlation coefficient (r^2) were systematically higher in corrected vs uncorrected studies.

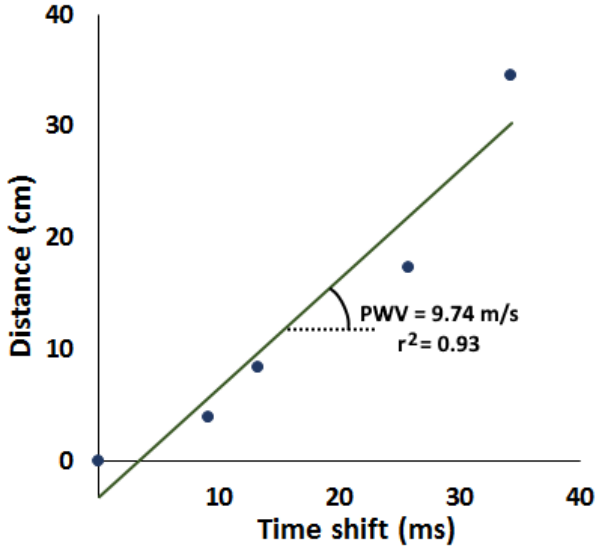

Fig. 2. Distance between the ROIs as a function of their corresponding time shifts (points in blue). In green is shown the regression line whose slope is the PWV.

TABLE I
CORRECTED AND UNCORRECTED VALUES OF PWV FOR EACH VOLUNTEER. THEY DID NOT SHOW SIGNIFICANT DIFFERENCES (P=0.23).

Uncorrected		Corrected	
PWV (m/s)	r^2	PWV (m/s)	r^2
9.74	0.93	10.00	0.94
11.00	0.52	17.67	0.69
12.00	0.35	11.93	0.58
9.62	0.84	10.02	0.86
12.15	0.67	11.83	0.74
9.00	0.83	8.91	0.87
14.13	0.60	12.81	0.70

IV. Discussion

In PC-MRI studies, the eddy currents cause distortions in the velocities that need to be corrected. These corrections were studied several years ago by Walker et al [13] for 2D+t studies, but the extrapolation to 4D Flow is still under investigation.

We proposed a method to estimate aortic PWV in 4D Flow MRI studies and analyze the difference generated by the zero correction. We acquired 4D Flow MRI sequences in seven volunteers and assessed their PWV. We compared the values with and without correction. We found that PWV does not change with the eddy currents correction. Zero errors modify the estimations of volumes and flows, where velocities are integrated through time. However, the computation of PWV is more sensitive to time-shifts than it is to amplitude differences.

In 4D Flow MRI studies a local value of PWV can be estimated, since the planes can be retrospectively and arbitrarily positioned. This is not the case in 2D+t studies, where the acquisition plane is defined prospectively. Nevertheless, 2D+t studies have higher spatial and temporal resolution and lower acquisition time than 4D Flow.

Currently, tonometry is still the standard method to assess PWV, despite its low accuracy in the estimation of distance.

This work had some limitations that need to be addressed. Namely, both the ROIs selection and the distance measurements were performed manually. To avoid intra and inter-observer variability, all measurements should be as automated as possible. However, although PWV is sensitive to distance assessment errors, the time-shifts are not as much affected to the ROI selection. We are aware that there are other simpler methods to estimate PWV, being tonometry the gold standard, whereas other methods are based on ambulatory pressure measurements (e.g. AASI) with known limitations reported elsewhere [16]. Advances in MRI processing are now more available to solve both the distance and temporal resolution issues.

V. Conclusion

In this work we presented a method to estimate PWV in 4D Flow MRI studies. We measured seven volunteers and analyzed the effect of eddy currents correction in the assessment of PWV. We found no significant differences in the values of PWV whether the correction was applied or not. Further studies should be conducted to incorporate PWV measurements in MRI studies as a complementary arterial stiffness indicator.

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