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MHD signal derived Auto Variable Velocity Encoding for 2D Flow Imaging in 3T Cardiac Magnetic Resonance Imaging

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***Abstract*—To develop a novel technique to set variable velocity-encoding (VENC) values according to magnetohydrodynamic (MHD) voltage/signal for 2D flow imaging in 3 Tesla MR system. MHD signal is calculated using the electrocardiogram signals measured outside and inside the static magnetic bore during the patient preparation process. Then, VENC values are assigned in terms of the MHD signal in each cardiac phase. A volunteer was scanned to evaluate the feasibility of the proposed method. Specifically, velocity and velocity to noise ratio (VNR) using the proposed method were measured and compared with conventional constant VENC value methods at 3T. MHD signal is measured during the patient preparation, thus no additional breath-holds are required and the VENC values can be calculated for each cardiac phase before the acquisition.**

I. INTRODUCTION

Phase contrast (PC) imaging is a well-established method in MRI to evaluate physiological properties of blood flow [1]. In a 2D prospective or retrospective gating PC sequence, bipolar gradients are generally applied to detect phase shift of moving spin, which is proportional to the velocity of the flow spin, thus enables the visualization of vascular structures during the cardiac cycle [2]. However, a velocity encoding (VENC) needs to be actively chosen before the acquisition in terms of the expected velocity of the vessel to be imaged. In general, PC sequence uses a single-VENC value to measure the flow velocity though entirely RR interval, which limits the measurement of blood flow velocity, since larger velocity-encoding values increase noise and smaller velocity-encoding values cause aliasing artefacts. Especially in pulsatile flow measurement such as the ascending aorta, a single-VENC value acquisition results in suboptimal velocity-to-noise ratio (VNR) during the diastole with a high VENC value or leads to the velocity aliasing during systole with a low VENC value. Thus, it is desirable to change the velocity encoding in a wide range during the RR interval.

To improve the accuracy and precision of pulsatile flow measurement, a method, using extra velocity encodings, has been proposed to provide additional information to unwrap phase aliasing and increase the VNR [3-5]. Buonocore et al firstly improved the measurement of blood flow in the RR interval by modifying the sequence and the raw data interpolation scheme to allow two VENC values during the RR interval [4]. However, only two different encodings, which are based on assumptions about the velocity range in the vessel of interest, have some limit for improving the accuracy through RR interval. More recently, triple-VENC flow imaging is proposed to acquire three datasets and then unwrap the aliased lowest VENC into a minimally aliased triple-VENC dataset without additional scan time [5]. However, a post-process algorithm is required to unwrap the phases, which probably induces unexpected phases or even expand the aliasing. Swan found that variable velocity encoding in a gated 2D PC sequence can better demonstrate signal intensity in large and small peripheral arterioles compared with a constant velocity-encoding sequence set to optimize signal at peak systolic flow [6,7]. Therefore, it is desirable to use variable VENC values to optimize the pulsatile flow measurement. But the challenge is how to set a profitable multiple velocity encoding values throughout the entire cardiac cycle. An automatic optimization of the VENC values was proposed by Ringgaard et al for individual cardiac phases in pulsatile flow according to a rapid low resolution pre scan [8]. The method significantly increases VNR in all phases by optimizing VENC values for each phase, however it comes at the expense of an additional scan time. Otherwise, the VENC is usually set according to the magneto-hydrodynamic (MHD) voltage, also named MHD signal. The MHD effect is a result of the static magnetic field and the displacement of moving electrical particles in the blood vessels [9]. Moreover, Ho Tse et al. found magneto-hydrodynamic in the static field is related to the flow velocity in aorta, and some groups have measured the beat-to-beat stroke volume across the MHD effect using a 12-lead ECG recording system [10-12].

In this paper, we propose a novel multiple VENC setting method without extra scans, in which the VENC values are set according to the MHD signal. Here, we firstly obtained an approximate flow variation by characterizing the MHD signal. Then, we analyzed the MHD signal and velocity

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retrospectively and used the MHD voltage to generate the flow estimation waveform. Finally the estimation waveform to dynamically adjust the flow VENC values through the entire cardiac cycle was validated in one subject.

II. THEORY AND METHODS

A. Theory of the MHD effect

Ions in blood plasma such as Na^+, Cl^- or HCO_3^- experience a Lorentz force under the B0 field in MR system. The Lorentz force depends on the magnitude and orientation of the blood flow velocity $\vec{v}$ with respect to the $\overrightarrow{B_0}$ field, which leads to the ions moving perpendicular to the direction of the blood flow and perpendicular to the B0 field. MHD effect thus occurs with the ions accumulating near the vessel's wall and leading to a potential difference across the vessel, which expresses as [13]:

$$V = \int_0^l \vec{v} \times \overrightarrow{B0}\, d\vec{l} \qquad (1)$$

Where $\vec{l}$ is the diameter of the vessel, V is the MHD voltage. The MHD voltage is proportional to the flow velocity. And the MHD effect alters the appearance of the ECG trace inside the MR scanner. More specifically, MHD signal superimposes on the real ECG signal inside the bore requires cardiac gating during cardiovascular MRI. ECG signal inside the bore can be expressed as the sum of real ECG signal and MHD signal:

$$ECG_{in} = ECG_{real} + VMHD \qquad (2)$$

The MHD effect results in ECG traces that are frequently unreadable due to the spectral peaks higher than the R-wave, which affects the ECG gating to erroneously trigger on MHD peaks. To improve the accuracy of real ECG in applications such as MRI gating and physiological monitoring, methods are proposed to eliminate the MHD signals [14,15]. However, basis of the relation of MHD signal and blood flow, the MHD signal is an alternative way to obtain noninvasive cardiac-output estimation, such as blood flow volume and stroke volume (SV). And thus, the MHD signal is used to derive the variable velocity encoding values in this work.

B. VENC Calculation

ECG traces are recorded during the registration process. Specifically, ECG_{out} trace of the subject is recorded with subjects laid on the patient table outside the bore. ECG_{in} trace is obtained with subjects laid on the patient table at the center of the bore. The recorded ECGs are processed offline using the MATLAB (The MathWorks, Natick, MA). More specifically, low pass filtering to remove high-frequency noise and interpolation to make sure the ECG traces at the same length are implemented. MHD signal is extracted by the subtraction of the processed ECG signals as:

$$\overline{VMHD} = \overline{ECG_{in}} - \overline{ECG_{out}} \qquad (3)$$

In Ref. [10], blood flow and left ventricular SV can be derived using MHD voltage extracted from intra-MRI ECG. Because the linear relation about flow and MHD as Eq. (1), the relative blood velocity v can be expressed as:

$$v = \frac{\overline{VMHD}}{d \times B_0} \qquad (4)$$

Where d is vessel width. VENC values are chosen for each cardiac phase according to this v through the entire cardiac cycle.

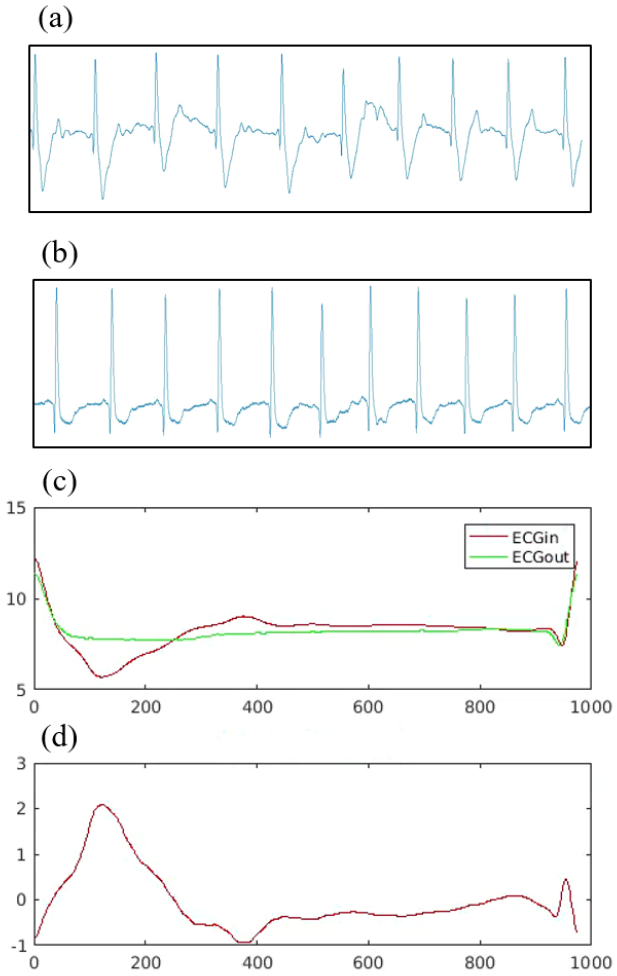

Figure 1. (a) Acquired ECG inside the MR scanner. (b) Acquired ECG outside the MR scanner. (c) Averaged ECG signals (d) Extracted MHD signal

C. Data Acquisition

The variable PC method was implemented with a 3 Tesla MR imaging system (uMR790, United Imaging Healthcare, Shanghai, China) equipped with maximum gradient amplitude 90 mT/m, slew rate of 200 mT/m/msec. A 4-lead ECG system (INVIVO) at a 1-kHz sampling rate only records physiological signals in the absence of gradient and radio frequency activity, removing the potential for variation in induced MHD voltage because of alteration in the MRI magnetic field. One healthy volunteer was studied with Institutional Review Board approval. ECG traces were obtained as described in section B. The subject received MHD signal derived auto dual-VENC scan (VENC1 = 140cm/s and VENC2 = 40cm/s), and two single-VENC scans with VENC = 140 cm/s and VENC=40cm/s individually. Other parameters were TR/TE = 14.1 ms / 4.36 ms, flip angle = 20° ; field of view = 280 × 320 mm^2; and slick thickness = 8 mm. The scan

was cardiac gated at 25 phases per cardiac cycle. The pulse sequence parameter VENC was expanded into an array with one element for each cardiac phase. To simplify the timing of the sequence, the echo time was kept constant for all cardiac phases. Because the gradient moment demand for the velocity-encoding segment increases with lower VENC values, the shortest possible echo time was defined by the cardiac phase with the lowest VENC value. VENC value for each cardiac phase could be input manually or loaded automatically from MHD calculation. All scans were reconstructed using k-t GRAPPA [16] with accelerated factor 2. All data underwent eddy current correction and concomitant Maxwell gradient correction before comparison.

D. In vivo flow analysis

Velocity noise and VNR were calculated using MATLAB (The MathWorks, Natick, MA). Velocity noise was evaluated by the flow-encoding velocity and the signal-to-noise ratio (SNR) of the magnitude images, when VNR was calculated by the velocity value to velocity noise.

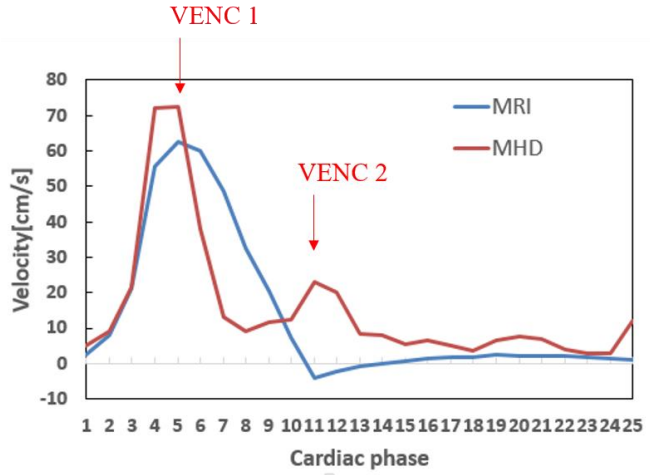

Figure 2. Comparison of averaged velocity of 2D PC-MRI and normalized MHD signal. Measurements were analyzed at two time points: VENC 1 for peak systole, and VENC 2 for early diastole before dicrotic notch peak.

III. RESULTS

Figure 1 shows MHD signal extraction procedure. ECG signals inside and outside the MR scanner were firstly recorded during the patient preparation. Then, ECG signals were interpolated to the same length and averaged according to the cardiac cycle length. After that, MHD signal was calculated by using Eq. (3). Figure 2 shows a MHD signal had been normalized and segmented to approximate averaged velocities throughout cardiac cycle of the artery, which was similar to the velocity curve from PC-MRI. VENC values were set as two times the velocity being arrowed in Figure 2.

Figure 3 demonstrates the average velocities in aorta measured with fixed single VENC=40cm/s (solid line in black), 140cm/s (solid line in blue) and MHD signal derived auto dual-VENC=[40,140]cm/s (solid line in red). Average VNR of single-VENC = 40 cm/s, single-VENC = 140 cm/s, and dual-VENC = [40,140] cm/s were 17.5, 16.6, and 20.2. Velocities were misestimated with low VENC value and phase wraps are shown in Figure 4(a), while velocities were correctly estimated with high VENC value in Figure 4(b) but with low VNR. MHD signal derived auto dual-VENC values setting combined the benefits of low and high VENC values methods, which shows the correct velocity with highest VNR in Figure 4(c).

IV. DISCUSSION AND CONCLUSION

In this study, we demonstrated an efficient auto multiple velocity encoding method for 2D PC MRI based on the MHD effect, which showed accuracy and precision in pulsation blood flow measurement. It has been found that MHD voltage could be used to estimate the subject's cardiac output in previous studies, thus we proposed MHD signal derived auto multiple velocity encoding for 2D PC MRI to improve the VNR and avoid phase wraps. The advantage of this method is that the VENC value calculation is completed during the patient preparation. However, the MHD effect mainly depends on the orientation of the static magnetic field with respect to the patient [17, 18]. Since that, we selected lead 1 to recording ECG signals due to its geometric orientation approximately perpendicular to both the B0 field and the aortic arch, which caused the strongest MHD effect.

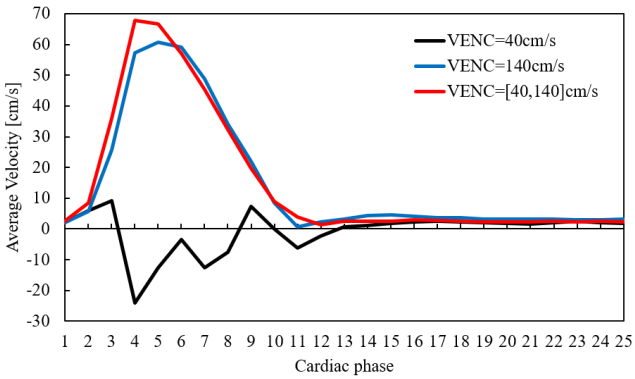

Figure 3. Average velocities in aorta measured with fixed single VENC = 40 cm/s (solid line in black), 140 cm/s (solid line in blue) and MHD signal derived auto dual-VENC = [40,140] cm/s (solid line in red).

The results show the MHD signal derived auto variable velocity encoding method can reliably assess the average velocity and help to set the VENC values for each cardiac phase in the aorta. The limitation of this work is that only one healthy subject was used to test the feasibility of MHD signal derived auto variable velocity encoding method. In the future, more human subjects would be included to validate the performance of the proposed method. Moreover, further validation is needful to explore whether the MHD signal derived auto variable velocity encoding is applicable for patient and other imaging techniques such as 4D flow imaging.

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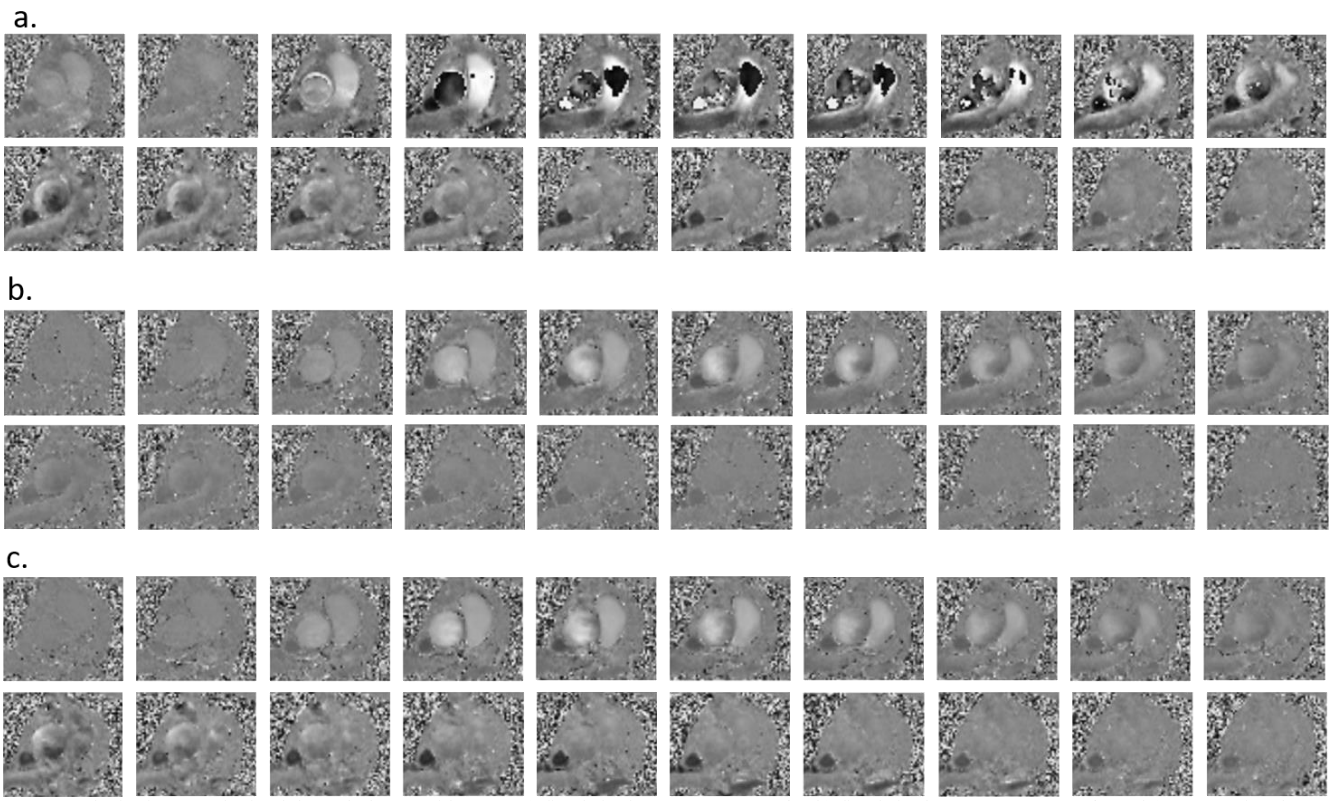

Figure 4. Velocity images obtained through the carotid artery.(a) fixed single-VENC = 40 cm/s, (b) fixed single-VENC = 140 cm/s, and (c) MHD signal derived dual-VENC =[40,140]cm/s. Average VNR of these three Velocity images are 17.5, 16.6, and 20.2.