Time Efficient Method for Dual-Venc Blood Velocity Measurements

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Introduction

Magnetic resonance angiography (MRA) can be used not only to get a morphological description of the blood circulation system but also to obtain additional diagnostic information such as the blood velocity distribution. To get this velocity information, a phase contrast (PC) MRA technique can be used. There are several trade-offs in the selection of parameters for PC velocity measurements. The accuracy of the velocity measurements is improved when strong velocity encoding (small Venc) is used. However, phase aliasing can corrupt the measurement if a wide range of velocities exist in the region of imaging. Thus, the Venc must be large enough to prevent aliasing yet as small as possible to maximize measurement accuracy. In cases when velocity distribution covers a wide range of velocities it is impossible to satisfy these requirements and simultaneously get high precision velocity measurements. In such situations, it is useful to acquire sets of images with 2 Vencs (low and high) [1]. Using 2 encoding velocities for each of 3 directions, plus a reference scan yields 7 complete image sets making the scan time for high-resolution 3D velocity imaging extremely long. We have developed a novel dual-Venc PC MRA method with improved time efficiency.

Theory and Methods

The proposed dual-Venc method is described in detail for the 1D velocity measurements case. The generalization to 2D or 3D cases is quite obvious.

Data Acquisition Scheme: Two images of the same slice are acquired in an interleaved mode. The first image is a reference or baseline image:

$$I_1(r) = I_0(r, v=0) + I_0(r, v\neq 0)$$

where v is a blood velocity, $I_0(r, v=0)$ are the image areas corresponding to the stationary tissues, and $I_0(r, v\neq 0)$ are the image areas corresponding to the flowing blood. The second image is acquired by applying velocity encoding gradient waveforms corresponding to two different Vencs: $Venc_1$ and $Venc_2$. Encoding with two Vencs is realized by the changing velocity encoding gradient amplitude. $Venc_1$ is used for odd k-space line acquisition and $Venc_2$ is applied during even k-space line acquisition. The resulting image is given by

$$I_2(r) = I_0(r, v=0) + I_A(r, v\neq 0) + I_B(r, v\neq 0)$$

$$I_A(r, v\neq 0) = I_0(r, v\neq 0)\cos(\frac{\phi_1-\phi_2}{2})\exp(i\frac{\phi_1+\phi_2}{2}) \qquad I_B(r, v\neq 0) = iI_0(r+\frac{FOV_y}{2}, v\neq 0)\sin(\frac{\phi_1-\phi_2}{2})\exp(i\frac{\phi_1+\phi_2}{2})$$

where FOV_y is field of view in the phase-encoding direction. In the case of constant velocity flow, the phases ϕ_1 and ϕ_2 are proportional to tissue velocity and inversely proportional to $Venc_1$ and $Venc_2$, respectively. The image $I_2(r)$ consists of original ($I_0(r, v=0)$ and $I_A(r, v\neq 0)$) and aliased ($I_B(r, v\neq 0)$) components.

Reconstruction Algorithm:

1. Two additional images ($I_3(r)$ and $I_4(r)$) are created from the acquired k-space data. Image 3 (4) is reconstructed from k-space data constructed in the following way: even (odd) k-space lines from the first image k-space data and odd (even) k-space lines from the second image k-space data.
2. Complex differences between $I_i(r)$ (i=2,3,4) and $I_1(r)$ are calculated:

$$\Delta I_{i1}(r) = I_i(r) - I_1(r)$$

3. Image $I_2(r)$ is dealised using the known relationships between the magnitudes of the complex differences $\Delta I_{i1}(r)$. The resulting images are the following:

$$I_{2A}(r) = I_0(r, v=0) + I_A(r, v\neq 0) \qquad I_{2B}(r) = I_B(r, v\neq 0)$$

4. Images $I_{2A}(r)$ and $I_{2B}(r)$ are combined to create two new images analogous to the completely sampled $Venc_1$ and $Venc_2$ encoded images:

$$I_{V1}(r) = I_0(r, v=0) + I_0(r, v\neq 0)\exp(i\phi_1) \qquad I_{V2}(r) = I_0(r, v=0) + I_0(r, v\neq 0)\exp(i\phi_2)$$

5. Phase difference images are calculated as usual

$$\phi_1(r) = PHASE(I_{V1}(r)/I_1(r)) \qquad \phi_2(r) = PHASE(I_{V2}(r)/I_1(r)) \text{ where } \phi_1 \text{ and } \phi_2 \in [-\pi, \pi]$$

Results

Phase contrast images of phantom and healthy volunteer were acquired on a 1.5T SIGNA Lx 8.4 scanner (GE Medical Systems, Milwaukee, WI) with NV/CVi gradients using the proposed technique. Figure 1 and 2 demonstrate the results of reconstruction from dual-Venc data.

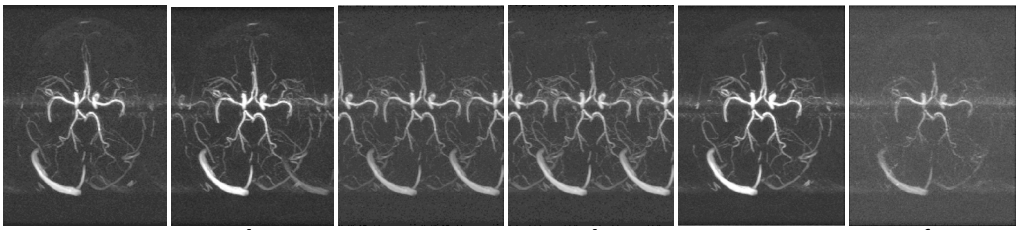

Figure 1. MIP of PC image volumes acquired by (**a**) the standard one-Venc and (**b-f**) the proposed dual-Venc techniques. **b**, **c**, and **d** calculated using compex differences between $I_i(r)$ $(i=2,3,4)$ and $I_1(r)$ show significant aliasing. Images **e** and **f** calculated using $I_{2A}(r)$ and $I_{2B}(r)$ demonstrate close to perfect dealising.

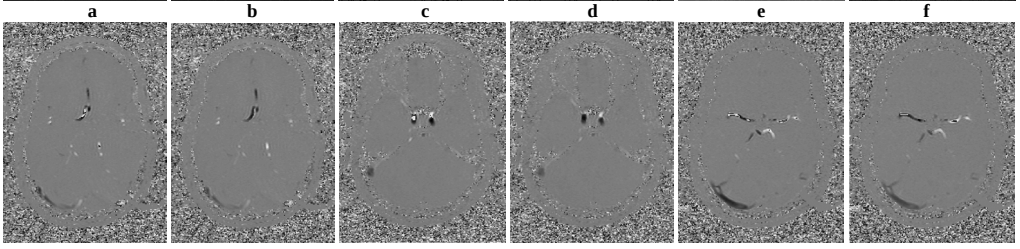

Figure 2 Phase difference images from dual-Venc data sets. Images with small Venc (**a**, **c**, and **e**) have significant phase (velocity) aliasing for vessels with high blood velocity. Images with larger Venc (**b**, **d**, and **f**) have noticeably reduced aliasing but small vessels are less visible than in small Venc images.

Discussion

A novel method to improve time efficiency of dual-Venc PC magnetic resonance velocity imaging has been developed. The proposed method requires only four velocity encoded measurements rather than seven measurements needed for the standard dual-Venc technique to reconstruct the complete 3D velocity map.

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References: [1] Lee AT, Pike GB, Pelc NJ. Magn Reson Med 1995;33:122-126.