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# A Novel Phase-Corrected 3D Cine Ultra-Short TE (UTE) Phase-Contrast MRI Technique

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*Abstract*—**Phase-contrast (PC) MRI is a non-invasive technique to assess cardiovascular blood flow. However, this technique is not accurate for instance at the carotid bifurcation due to turbulent and disturbed blood flow in atherosclerotic disease. Flow quantification using conventional PC MRI distal to stenotic vessels suffers from intravoxel dephasing and flow artifacts. Previous studies have shown that short echo time (TE) potentially decreases the phase errors. In this work, a novel 3D cine UTE-PC imaging method is designed to measure the blood velocity in the carotid bifurcation using a UTE center-out radial trajectory and short TE time compared to standard PC MRI sequences. With a new phase error correction technique based on autocorrelation method, the proposed 3D cine UTE-PC has the potential to achieve high accuracy for quantification and visualization of velocity jet distal to a stenosis. Herein, we test the feasibility of the method in determining accurate flow waveforms in normal volunteers.**

## I. INTRODUCTION

Phase-contrast (PC) MRI is a reliable non-invasive technique for characterization and quantification of cardiovascular blood flow [1,2]. However, this technique is not accurate in cases where there is atherosclerotic disease or where the blood flow is disturbed or turbulent. Disturbed flow may be seen at bifurcations, branch point, and other regions of the arterial tree where the blood flow is altered. These areas are more prone to atherosclerosis and narrowing in vessels, leading to turbulent blood flow and a high velocity jet distal to a stenosis [3-6]. A signal loss may appear distal to a stenosis due to this turbulent blood flow.

In addition, turbulent flow results in velocity fluctuations, leading to intravoxel dephasing and significant error in velocity measurement and assessment of blood flow. Carotid bifurcation is one of the main sites of atherosclerosis and is a good example of complex and disturbed blood flow due to atypical geometry of this branch site [7].

Several approaches have been developed to potentially reduce the signal loss in PC-MRI images [3,8-11]. However, a reliable flow measurement technique in the presence of turbulence remains elusive. One important approach that revealed significant impact in correction of the signal loss involves reduction of the echo time (TE) and gradient duration [12-15]. Reducing the TE decreases the impact of turbulent fluctuation velocity, intravoxel dephasing and the subsequent signal loss. The approach results in higher signal to noise ratio and more reliable estimation of disturbed and jet flows in PC MRI since a shorter TE will ameliorate the effect of intravoxel dephasing caused by random fluid mixing.

The 2D UTE-PC technique has previously been investigated for measurement of blood flow through the aortic valve [16]. UTE techniques in general suffer from phase errors due hardware imperfections, specifically gradient channel delays. This problem becomes quite pronounced in flow measurement wherein the phase information of image needs to be used for quantification of flow. In addition to the inherent limitations of the UTE technique, the previously proposed 2D UTE-PC technique by O'Brien et al. has other issues of concern. The higher sensitivity of technique to scanner imperfection, specifically $B_0$ inhomogeneity, can cause phase errors and velocity miscalculation. In addition, the combination of flow encoding gradient and slice selection gradient in this technique results in dependency of slice selection gradient on the velocity encoding gradient. This leads to limitation in minimum slice thickness as well as dependency of slice thickness, velocity-encoded gradient strength and velocity encoding (Venc). With this dependency, flow assessment becomes only possible in the through-plane direction.

In this work, we present a novel 3D cine UTE-PC sequence to obviate the problems with previous UTE-PC technique. We also show that the correction of phase errors produced by gradient delays dramatically improves the result of the UTE-PC technique. The proposed sequence is utilized to quantitatively measure blood velocity in the carotid bifurcation which is associated with disturbed blood flow and results are compared to standard PC MRI sequences.

## II. MATERIAL AND METHOD

### A. *Pulse sequence*

Phase-contrast MRI is based on phase differences of flowing spins compared to stationary spins. Generally a flow sensitive (encoding) and a flow compensated (reference) scan are acquired by applying a bipolar gradient in the direction of interest. By subtracting the phase images of the two scans, a flow image is achieved. Figure 1(a) demonstrates a standard PC MRI sequence using Cartesian trajectory. In this sequence slice excitation, refocusing, and flow encoding gradients need to be applied during TE. These gradients which are as long as the duration of TE (TE is defined as the distance between the center of the RF pulse

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to the center of the readout window), prolong the TE, leading to phase errors and velocity miscalculation.

The previously proposed 2-D UTE technique [16] shortened the TE through two approaches: (i) by combining the slice selection gradient with flow encoding/compensation gradient and (ii) by starting data acquisition from the beginning of gradient ramp using non-linear sampling. TE is defined in this case as the distance between the center of the RF pulse to the beginning of the readout gradient. The strategy results in few potential problems. The inherent velocity sensitivity of slice select gradients and inverting the sign of the slice select gradient, permits a shorter combination of the slice select gradient with the flow encoding gradient and achieving a shorter TE. Inverting the sign of slice select gradient, in general causes more sensitivity to scanner imperfections, and in particular, $B_0$ inhomogeneities resulting in phase errors and velocity miscalculations. In addition, combining the flow encoding and slice select gradients leads to dependency of slice select gradient on the velocity encoding gradient. This dependency causes limitations on the minimum slice thickness and therefore the velocity-encoded gradient strength which is controlled by the velocity encoding (Venc) parameter. The minimum achieved TE with the technique reported in [16] was 0.65 msec with Venc=500 cm/s and a minimum slice thickness of 8.75 mm which often is not clinically applicable. On top of that, with this dependency, flow assessment is only possible in the through-plane direction

Figure 1(b) shows the proposed 3D UTE-PC sequence diagram where the slice excitation gradient is combined with flow encoding/compensation gradient. As with the previous 2-D technique, k-space sampling starts with the rising slope of readout gradient. Two back-to-back RF pulse and readout gradients are used in order to construct one flow sensitive image and one flow compensated (reference) image. In comparison to [16] the sign of slice excitation gradient is not inverted (at the cost of slightly longer TE) and therefore the high sensitivity of previous technique to hardware imperfection is resolved. In addition, only the slice refocusing section of the slice excitation gradient is combined with the flow encoding gradient which results in less dependency of flow encoding and slice thickness with the penalty of a slightly longer TE. The minimum TE for Venc=500 and slice thickness 8.75mm (identical to previous technique) is 0.76 msec. However, the slice thickness in this technique can be significantly reduced.

The radial readout trajectory in UTE is based on radial traversal of evenly spaced k-space lines starting from the center of k-space and ending on the surface of a cylinder with radius $K_{max}$ determined by spatial resolution in the in-plane direction. A 3D non-Cartesian trajectory acquisition based on the “stack-of-stars” [17] strategy was employed to collect multiple slices in a cylindrical volume in the flow encoding direction.

### B. *Phase error correction*

Center-out k-space radial trajectories help to reduce the effect of intravoxel dephasing and related phase artifacts due to inherent first moment minimization of the readout

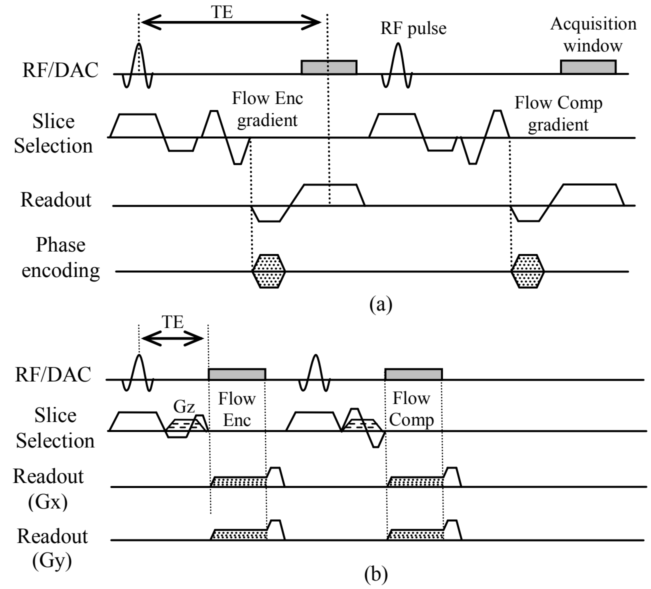


Figure 1. Conventional 2D PC MRI sequence (a) and proposed 3D UTE-PC MRI sequence (b). In the proposed sequence, a flow sensitive and a flow compensated scan are acquired through combination of a bipolar gradient and refocusing part of slice select gradient. Gradient $G_z$ is applied to acquire multiple slices in a volume using stack of star strategy.

gradient by oversampling the center of k-space [1]. However, center-out k-space lines are sensitive to phase errors due to several parameters such as gradient delays, eddy currents, and $B_0$ field inhomogeneity. This can result in a mis-centering of k-space trajectory and lead to image artifact as well as erroneous phase calculations. To resolve the phase error in the proposed UTE technique with center-out radial readout, a phase error correction technique has been proposed based on auto-correlation [18].

Since the gradient delays in three physical gradient directions are independent, the consequent k-space mis-centering and phase errors in each three directions are independent as well. Assuming regridding of radial k-space to a rectilinear k-space, and three independent phase errors in three k-space directions, the phase distorted MR image $f_1(x,y,z)$ can be expressed as:

$$f_1(x,y,z)=f(x,y,z)e^{i(x.\Delta\phi_x+y.\Delta\phi_y+z.\Delta\phi_z)} \qquad (1)$$

where f, $\Delta\phi_x$, $\Delta\phi_y$, and $\Delta\phi_z$ represent the original image matrix without phase error and phase error in x, y, and z directions in the measured image, respectively. Considering the reconstructed data as a 3-D volume with M×N×W voxels, matrix f has the same number of elements. By repeating the same scan with inverted readout gradients, the phase errors in three directions will be inverted. Thus, the reconstructed image from the second scan is

$$f_2(x,y,z)=f(x,y,z)e^{-i(x.\Delta\phi_x+y.\Delta\phi_y+z.\Delta\phi_z)} \qquad (2)$$

Considering independent phase error in three directions, normalized cross-correlations between two scans can be performed separately in each direction using M×N×W matrices $f_1$ and $f_2$. To calculate the cross-correlation, matrix $C_z$ can be written as

$$C_z = \sum_m f_1(x,y,m)\, f_2^*(x,y,m) \quad (3)$$

Assuming phase error only in z direction, if we substitute (1) and (2) in (3), we find

$$C_z = \sum_m f(x,y,m)\, f^*(x,y,m) e^{-i2(z.\Delta\phi_z)} \quad (4)$$

where $C_z$ is the M×N correlation matrix between two scan in z direction. Normalized cross-correlation between the two scans in z direction can be expressed as

$$R_z = \frac{C_z}{|C_z|}$$

$$R_z = \frac{\sum_m f(x,y,m)\, f^*(x,y,m) e^{-i2(z.\Delta\phi_z)}}{\left|\sum_m f(x,y,m)\, f^*(x,y,m)\right|} \quad (5)$$

Since $f(x,y,z)\, f^*(x,y,z)$ is positive and real valued for MR images and its phase is always zero, phase shift in z directions can be measured as

$$\arg(R_z) = \arg\left(\frac{\sum_m f(x,y,m)\, f^*(x,y,m) e^{-i2(z.\Delta\phi_z)}}{\left|\sum_m f(x,y,m)\, f^*(x,y,m)\right|}\right) \quad (6)$$

$$\arg(R_z) = 2z.\Delta\phi_z$$

$$\Delta\phi_z = \frac{1}{2z}.\arg(R_z) \quad (7)$$

From equation (7), it can be seen that the phase shift in z direction is proportional to the phase of normalized cross-correlation of two scans. Alternately, phase shift in x and y directions, $\Delta\phi_x$ and $\Delta\phi_y$, can be calculated using

$$\Delta\phi_x = \frac{1}{2x}.\arg(R_x) \quad (8)$$

$$\Delta\phi_y = \frac{1}{2y}.\arg(R_y) \quad (9)$$

where $R_x$ and $R_y$ are

$$R_x = \frac{\sum_k f(k,y,z)\, f^*(k,y,z) e^{-i2(x.\Delta\phi_x)}}{\left|\sum_k f(k,y,z)\, f^*(k,y,z)\right|} \quad (10)$$

$$R_y = \frac{\sum_j f(x,j,z)\, f^*(x,j,z) e^{-i2(y.\Delta\phi_y)}}{\left|\sum_j f(x,j,z)\, f^*(x,j,z)\right|} \quad (11)$$

By measuring the phase error in each direction the phase corrected image will be

$$f(x,y,z) = f_1(x,y,z) e^{-i(x.\Delta\phi_x + y.\Delta\phi_y + z.\Delta\phi_z)} \quad (12)$$

Based on the Fourier shift theorem, equation (12) can be written as

$$F(k_x,k_y,k_z) = F_1(k_x+\Delta k_x, k_y+\Delta k_y, k_z+\Delta k_z) \quad (13)$$

where FT and $FT_1$ are Fourier transform of images f and $f_1$ in k-space domain. $\Delta k_x$, $\Delta k_y$, and $\Delta k_z$ are trajectory delays in $k_x$, $k_y$, and $k_z$ directions which they are equal to $\Delta\phi_x/2\pi$, $\Delta\phi_y/2\pi$, and $\Delta\phi_z/2\pi$ respectively.

### C. *Imaging strategy*

In order to reduce the phase error, a one-time calibration is performed by scanning a static phantom prior to the in-vivo scan based on the above phase correction technique.

TABLE I. ACQUISITION PARAMETER FOR CARTESIAN PC MRI AND UTE-PC MRI USING STACK OF STAR TRAJECTORIES

| Parameter | Cartesian PC MRI | UTE-PC MRI |
|---|---|---|
| FOV (mm) | 160x188x5 | 170x170x50 |
| TE (ms) | 2.9 | 1.08 |
| TR (ms) | 4.9 | 6.2 |
| spatial resolution(mm) | 2.0x2.0x5.0 | 1.17x1.17x5 |
| Flip angle (deg.) | 15 | 15 |
| Venc (cm/s) | 200 in through plane direction | 200 in through plane direction |

The measured trajectory delays using the proposed phase correction technique for each direction were subtracted from each k-space trajectory. These trajectory delays were then applied to all subsequent in-vivo scans since scan parameters for the static phantom and in-vivo studies are identical.

Imaging was performed on an Achieva 1.5T Philips scanner using a combined 16-element SENSE Neurovascular coil capable of imaging carotid vessels from the aortic arch to the Circle of Willis.

For the in-vivo study, four normal volunteers with an average of 27±4 years of age were scanned using standard multi-2D cine PC MRI sequences as well as the proposed 3D cine UTE-PC MRI sequence. Imaging was performed perpendicular to the common carotid artery covering an axial 5cm 3D volume including 10 slices with 5mm slice thickness. The other parameters for the conventional and proposed technique are shown in table I. The multi-2D and 3D volume was located 1 cm proximal to 4 cm distal to the carotid bifurcation. Blood flow was evaluated in the right and left common carotid artery (CCA) proximal to carotid bifurcation.

## III. RESULTS AND DISCUSSIONS

Figure 2 demonstrates the results from phantom study using proposed 3D UTE-PC sequence with stack-of-stars trajectory in three flow directions. The top row shows the three images with different flow directions without trajectory delay correction. The bottom row shows the images after calculation of trajectory delay using phase-shift auto-correlation method and their subsequent application for

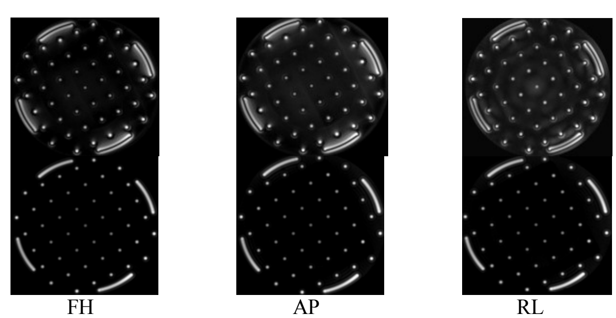


Figure 2. Phantom result for different flow directions. The first row shows the original images using UTE-PC imaging in FH, AP, and RL flow direction respectively. The second row shows the result after mis-centered k-space correction using trajectory delays of 2, 2.5, and 4 for FH, AP, and RL flow directions.

TABLE II. COMPARISON OF FLOW MEASUREMENT BETWEEN 3D UTE AND CONVENTIONAL PC MRI IN 4 NORMAL VOLUNTEERS. QUANTITIES HAVE BEEN AVERAGED FOR THE RIGHT CCA.

| | Multi-2D Cart PC MRI | 3-D UTE-PC MRI without phase corr. | | 3-D UTE-PC MRI with phase corr. | |
|---|---|---|---|---|---|
| | | | Difference from cart PC MRI (%) | | Difference from cart PC MRI (%) |
| Peak systolic velocity (cm/s) | 72.6±11 | 87.2±14 | 20.1 | 75.3±7 | 3.7 |
| Mean systolic velocity (cm/s) | 50.3±13 | 63.7± 13 | 26.6 | 54.3±11 | 7.9 |
| Peak systolic flow(mL/s) | 17.8±7 | 21.5± 8 | 20.8 | 19.1±6 | 7.3 |

each direction. The images after trajectory delay correction reveal significantly less phase error resembling ghosting in first row images. This phase error has considerable effect in phase-contrast technique and flow quantification wherein phase variation in flowing blood is desirable and any other source of phase deviation should be eliminated.

The effect of phase error correction was also investigated in an in-vivo study on a normal volunteer. Figure 3 demonstrates the peak velocity, mean velocity and flow in right CCA 0.5 cm proximal to bifurcation using conventional PC MRI and proposed 3D UTE-PC MRI with and without trajectory delay correction. Flow waveforms and temporal evolution for peak and mean velocity for UTE-PC MRI after phase error correction have an excellent correlation with conventional technique and the differences between the peak systolic velocity using conventional and UTE-PC sequence after phase error correction is less than 4%. However, this difference in case of UTE-PC MRI without phase error correction is more than 10%. In addition, magnitude and phase images for UTE-PC with phase error correction show less artifact compared to UTE-PC without phase error correction.

The result of blood velocity measurement in right CCA for one slice located at 0.5 cm proximal to bifurcation for all volunteers using standard multi-2D PC MRI and 3D UTE-PC MRI before and after phase error correction is summarized in table II. The last column in table II represents the difference of measured quantities between conventional PC MRI and 3D UTE-PC MRI after phase error correction. The velocity and flow measurements for 3D UTE-PC MRI after phase error correction are different from conventional PC MRI by less than 8% while this difference falls to 20% when there is no phase error correction. In addition, the flow measurement were performed for left CCA 0.5 cm proximal to carotid bifurcation as well as for right and left internal carotid artery (ICA) and external carotid artery (ECA) 1 cm distal to carotid bifurcation and the results were similar to the demonstrated measurement for CCA in table II. This proves our claim that proposed sequence lead to acceptable flow quantification with respect to conventional technique in normal volunteer. The TE in the proposed technique is shorter than conventional technique which could lead to more accurate measurement in the case of vascular stenosis.

## IV. CONCLUSIONS

The present study demonstrates that a 3D cine UTE-PC MRI results in reasonable flow quantification in normal volunteers with respect to conventional multi-2D cine PC Previous phantom study on 2D UTE-PC MRI revealed 27% underestimation of flow rate compared to the standard technique at Q=100 mL/s (Reynolds number=7200) [16].

This error will be even more severe at lower flow rates. This is most likely due to high sensitivity of the technique to system imperfections which lead to phase errors with the UTE technique. In [21], we have shown that the accuracy of proposed 3D UTE technique is comparable with standard PC MRI technique in the setting of flow rate as low as 13.2 mL/s (Reynolds number=160). This improvement is likely

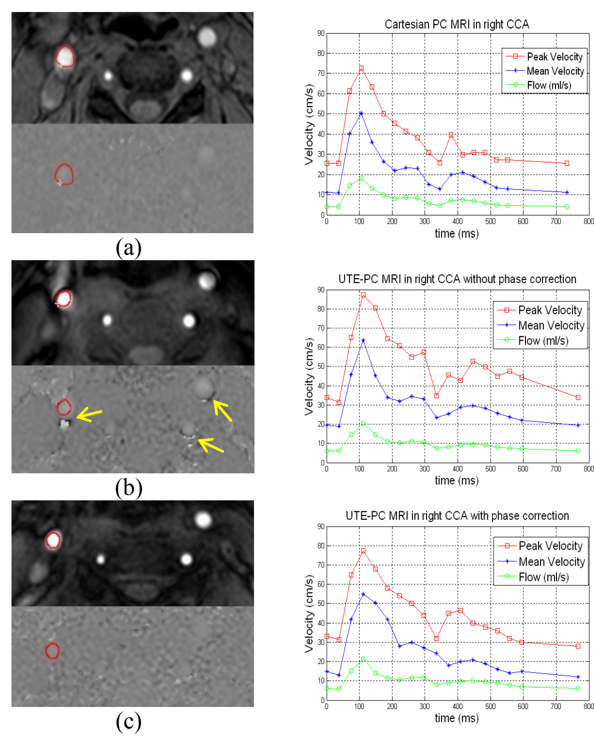


Figure 3. Right CCA magnitude and phase image (left column) and peak velocity, mean velocity and flow waveform (right column) for conventional sequence (a) UTE-PC MRI without phase error correction (b) and UTE-PC MRI with phase error correction (c). Image slice from the M2D acquisition in (a) and 3D acquisition in (b) and (c) was located 5 mm proximal to bifurcation.

due to the phase correction step as well as less phase error due to eddy current in proposed 3D UTE technique. Further we have shown the accuracy of flow measurement is higher even for the case of low flow rates in carotid arteries. Unlike conventional 2D UTE MRI techniques, the current method can be tailored to significantly reduce the slice thickness making it a robust technique applicable to many different clinical applications. In addition, by resolving the dependency of slice thickness and flow encoding gradients of previous UTE techniques, determination of components of in-plane flow will become feasible.

The scan time for 3-D UTE-PC MRI sequence is longer than conventional Cartesian PC MRI. This is due to the required $\pi\times N\times N$ k-space lines to cover the whole k-space (with the Cartesian technique $N\times N$ k-space lines are sufficient and for radial $\frac{\pi}{2}\times N\times N$ are needed). It is therefore crucial to speedup image acquisition for 3-D UTE. One approach to scan-time reduction is to under-sample the radial projections. However, under-sampling the k-space can cause streaking artifacts and reduced SNR. These issues will be investigated as part of future work.

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