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3-D and 4-D phase unwrapping methods applied to phase contrast magnetic resonance velocity imaging

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ABSTRACT

Phase contrast magnetic resonance velocity imaging is an established technique used for quantitative blood flow measurements. The velocity encoding sensitivity parameter V_{enc} (velocity for a phase shift of π) is usually set above the maximum blood velocity V_{max} to prevent the phase from being wrapped modulo 2π. As a result, dynamic range and signal-to-noise ratio are sacrificed. In this paper the performance of three different phase unwrapping algorithms on datasets obtained with a range of V_{enc} values below V_{max} is investigated. A simple temporal unwrapping algorithm was found to fail at V_{enc} below $0.6V_{max}$. A full path-independent three-dimensional algorithm, unwrapped successfully at V_{enc} values down to $0.4V_{max}$. Finally a novel four-dimensional algorithm is proposed which involves unwrapping along the velocity encoding direction, and which was found to be successful at V_{enc} values as low as $0.2V_{max}$. Compared to the traditional approach, both dynamic range and signal-to-noise ratio are thereby increased by a factor of up to five times.

1. INTRODUCTION

Phase contrast magnetic resonance imaging (PC-MRI) produces a linear relationship between the velocity of blood v and the phase shift of the magnetic resonance signal ϕ:

$$v(\phi) = \frac{\phi \cdot V_{enc}}{\pi}. \qquad (1)$$

The constant V_{enc} is called the velocity encoding parameter and is equal to the velocity that results in a phase shift of π radians [1]. When V_{enc} is smaller than the maximum blood velocity, V_{max}, the phase is wrapped onto the range $[-\pi,\pi)$. The standard procedure is to avoid the phase wrapping problem by choosing V_{enc} values equal to or larger than V_{max}. However, this may compromise the accurate measurement of the lower velocities close to the walls of the arteries, an important requirement for measuring wall shear stresses. Therefore, the use of lower encoding velocities can help to improve the dynamic range and signal to noise ratio of the images, making them more suitable for wall shear stress measurements.

When PC-MRI is combined with cardiac triggering a series of two-dimensional (2-D) phase contrast MR images of a single slice are produced to form a three-dimensional (two spatial axis and one time axis) phase volume. Unwrapping a three-dimensional (3-D) phase volume by any of the existing 2-D methods frame by frame requires correction of the resulting phase

offset between frames. By contrast, taking account of the 3-D nature of the data can be an advantage to improve the unwrapping process. However, there is only a small number of published papers on the 3-D phase unwrapping problem [2-4].

2. MATERIALS AND METHODS

Phase-contrast axial CINE MRI acquisitions of a single subject, through the ascending aorta, were obtained with varying velocity encoding values of 25, 50, 75, 100, 125, 150 cm s^{-1}. The maximum blood velocity was V_{max} = 120 cm s^{-1}. Three different unwrapping methods were compared: a noise-immune three-dimensional phase unwrapping algorithm [2], a temporal unwrapping method [5], and a novel four-dimensional (4-D) unwrapping method based on applying temporal unwrapping with encoding velocity instead of time as the unwrapping direction.

2.1 Three-dimensional Phase Unwrapping

The 3-D phase unwrapping method consists of the following steps [6]. First, residues are identified in the phase volume, and the type, sign and location of each residue are stored. Then, the residues are sorted to form loops. Although residues generally form closed loops, open phase singularity loops ending on the phase boundary, also called partial loops, can be present. A partial loop is closed by connecting its ends with artificial residues added on the phase volume boundary. After all loops were identified and closed, the branch surface for each was set by shrinking the loop towards its geometric centre and in this process flags were set up to indicate that the replaced segments constituted a forbidden unwrapping path. Once all the flags indicating where the branch surfaces were located had been set, the unwrapping was carried out with a flood fill algorithm.

2.2 Temporal Phase Unwrapping

The temporal unwrapping algorithm takes as input a set of two-dimensional wrapped phase maps, $\phi_w(x, y, t)$ where (x, y) refers to the pixel location and t represents non-dimensional time (t = 0,1,..s). Subscripts w and u denote wrapped and unwrapped phases, respectively. The phase change at a given pixel between times i and j is given by

$$\Delta\phi(i, j) = \phi(i) - \phi(j) \tag{2}$$

where the pixel coordinates have been dropped for clarity. This phase change is wrapped onto the range ($-\pi$, π) by the unwrapping operator U:

$$\Delta\phi_w(i, j) = U\{\Delta\phi(i, j), 0\} \tag{3}$$

$U\{\phi_1, \phi_2\}$ subtracts an integral multiple of 2π from ϕ_1 such that $\phi_1 - \phi_2$ lies in the range $-\pi$ to $+\pi$:

$$U\{\phi_1, \phi_2\} = \phi_1 - 2\pi\text{NINT}[(\phi_1 - \phi_2)/2\pi] \tag{4}$$

where NINT[...] denotes rounding to the nearest integer.
The total unwrapped phase change at time t is calculated by summing the wrapped phase differences:

$$\phi_u(t) = \sum_{t'=1}^{t} \Delta\phi_w(t', t'-1) \tag{5}$$

where $\phi(0)$ is defined to be equal to 0.

2.3 Four-dimensional Phase Unwrapping

The 4-D unwrapping is based on using the temporal phase unwrapping method with velocity encoding instead of time as the unwrapping direction. This approach makes use of four dimensions: *x, y, t* and V_{enc}. Each voxel (*x, y, t*) is unwrapped independently of the rest of the voxels using the velocity encoding dimension.

Given that there is a linear relationship between the measured phase and the inverse of the velocity encoding, it is possible to use an exponential sequence of V_{enc}^{-1} values [7], which allows the unwrapping technique to be used with a reduced amount of data. A growing exponential sequence of V_{enc}^{-1} values of 0.005, 0.01, 0.02, and 0.04 (i.e. encoding velocities values equal to 200, 100, 50 and 25 cm s^{-1}) was used to unwrap each voxel. The analogue of time here is $g = V_{enc}^{\max}/V_{enc}$ where $V_{enc}^{\max}$ is the maximum value of V_{enc} used in the sequence and which must lie above the maximum blood flow velocity to ensure no wraps occur. Since the volume with V_{enc} = 200 cm s^{-1} had not been acquired, it was simulated by re-scaling the 150 cm s^{-1} volume. With these four points, the unwrapping proceeds as follows. The wrapped increment $\Delta\phi_w(2,1)$ is unwrapped using the phase value $\phi_w(1)$:

$$\Delta\phi_u(2,1) = U\{\Delta\phi_w(2,1), \phi_w(1)\}. \tag{6}$$

The spatial and true temporal indices have been dropped from Eq. (6) for reasons of clarity. $\Delta\phi_u(2,0)$ is calculated as

$$\Delta\phi_u(2,0) = \Delta\phi_u(2,1) + \phi_w(1). \tag{7}$$

$\Delta\phi_u(2,0)$ can now be used to unwrap $\Delta\phi_w(4,2)$:

$$\Delta\phi_u(4,2) = U\{\Delta\phi_w(4,2), \Delta\phi_u(2,0)\} \tag{8}$$

from which $\Delta\phi_u(4,0)$ is calculated as:

$$\Delta\phi_u(4,0) = \Delta\phi_u(4,2) + \Delta\phi_u(2,0). \tag{9}$$

This process is repeated using the phase values $\phi(g)$ (g = 1,2,4,8,...s). Eqs.(8) and (9) can be rewritten for the general case:

$$\Delta\phi_u(2g, g) = U\{\Delta\phi_w(2g, g), \Delta\phi_u(g,0)\} \tag{10}$$

$$\Delta\phi_u(2g,0) = \Delta\phi_u(2g, g) + \Delta\phi_u(g,0) \tag{11}$$

3. RESULTS AND CONCLUSIONS

Temporal and spatial undersampling of the data limited the performance of the 3-D and temporal unwrapping methods. The temporal unwrapping algorithm unwrapped successfully V_{enc} values above $0.6V_{max}$ (Figs. 1 and 2), while the 3-D algorithm unwrapped successfully V_{enc} values above $0.4V_{max}$ (Figs. 3 and 4), and the 4-D algorithm succeeded in unwrapping the lowest V_{enc} equal to $0.2V_{max}$ (Figs. 5 and 6). Successful unwrapping of phase contrast images obtained with low velocity encoding results in improved velocity to noise ratio. This can be appreciated in the velocity surface inside the ascending aorta obtained with V_{enc} = 150 cm s^{-1} (Fig. 7) and V_{enc} = 25 cm s^{-1} (Fig. 8). This improvement is quantified in Fig. 9 by calculating the rms velocity fluctuations across the ascending aorta at each V_{enc} value at a point in the cardiac cycle where the velocity field is approximately zero. Compared to the approach of having $V_{enc} \geq V_{max}$, both dynamic range and signal-to-noise ratio can be increased by a factor of up to five times when a lower V_{enc} is used.

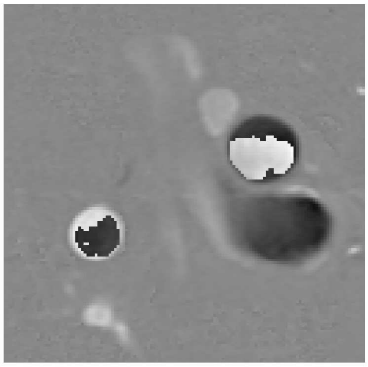

Fig.1. Wrapped phase corresponding to highest velocity frame, with V_{enc} = 75 cm s^{-1}.

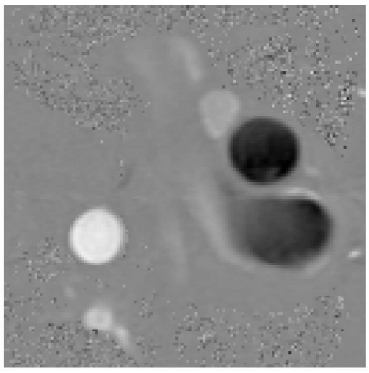

Fig. 2. Unwrapped phase corresponding to Fig. 1, using temporal unwrapping method.

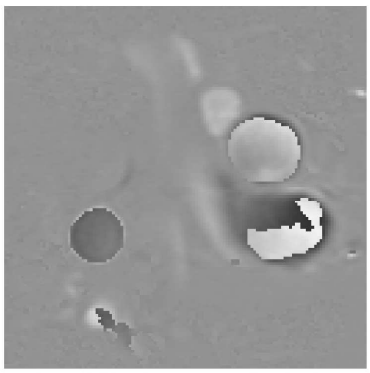

Fig.3. Wrapped phase corresponding to highest velocity frame, with V_{enc} = 50 cm s^{-1}.

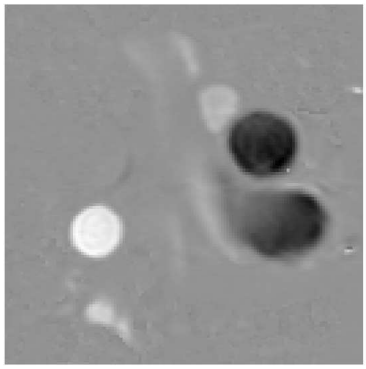

Fig. 4. Unwrapped phase corresponding to Fig. 1, using 3-D unwrapping method.

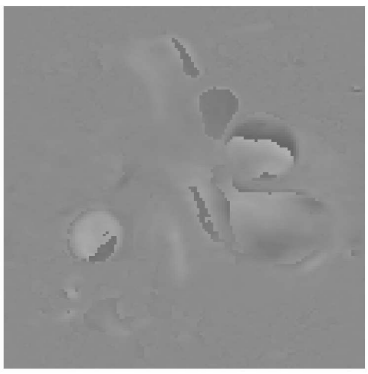

Fig.5. Wrapped phase corresponding to highest velocity frame, with V_{enc} = 25 cm s^{-1}.

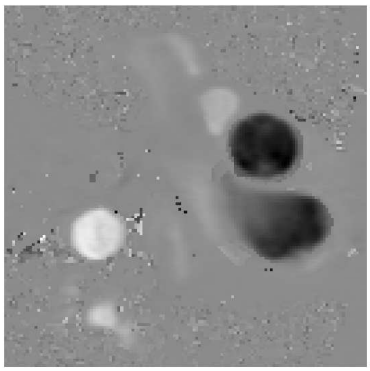

Fig. 6. Unwrapped phase corresponding to Fig. 1, using 4-D unwrapping method.

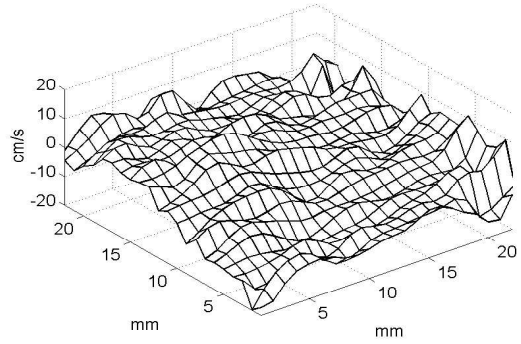

Fig. 7. Velocity distribution inside the ascending aorta obtained with V_{enc}=150 cm s^{-1}.

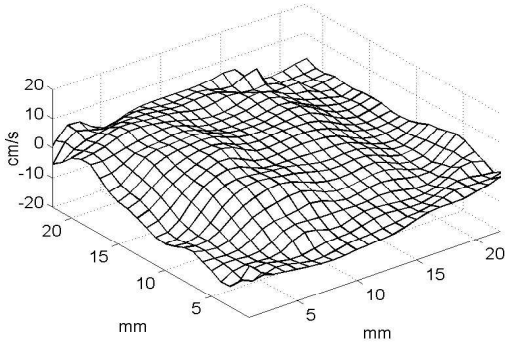

Fig. 8. Velocity distribution inside the ascending aorta obtained with V_{enc}= 25 cm s^{-1}.

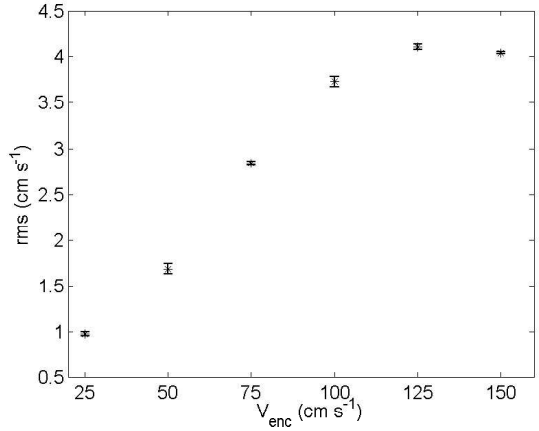

Fig.9. Rms velocity fluctuations across the ascending aorta versus V_{enc} when the velocity field is approximately zero

BIBLIOGRAPHY

[1] D. W. McRobbie, E. A. Moore, M. J. Graves, and M. R. Prince, MRI from picture to proton, Cambridge University Press, Cambridge, UK, 2003

[2] J. M. Huntley, "Three-dimensional noise-immune phase unwrapping algorithm", *Appl. Opt*. Vol. 40, 2001, pp. 3901-3908

[3] R. Cusack and N. Papadakis, " New robust 3D phase unwrapping algorithms: application to magnetic field mapping and undistorting echoplanar images", *NeuroImage*. Vol. 16, 2002, pp. 764-764
[4] M. Jenkinson, "Fast, automated, N-dimensional phase-unwrapping algorithm", *Magn. Reson. Med.* Vol. 49, 2003, pp. 193-197
[5] J. M. Huntley and H. O. Saldner, "Shape measurement by temporal phase unwrapping: comparison of unwrapping algorithms", *Meas. Sci. Technol.*Vol. 8, 1997, pp. 986-992
[6] O. Marklund, J. M. Huntley, and R. Cusack, "Robust unwrapping algorithm for 3-D phase volumes of arbitrary shape containing knotted phase singularity loops", *IEEE Trans. Med. Imag.*, to be published
[7] J. M. Huntley and H. O. Saldner, "Shape measurement by temporal phase unwrapping: comparison of unwrapping algorithms", *Meas. Sci. Technol.*, Vol. 8, 1997, pp. 986-992