Velocity Contour Mapping For Rapid Practical Flow Examination

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PURPOSE

In fluid dynamics, 2D distributions are conveniently visualized using contour maps. Here, we describe an MRI acquisition technique that directly encodes velocity isocontours in the signal amplitude without post-processing. This method provides a fast and convenient way to map velocity in MRI.

THEORY

The proposed excitation (Fig 1) is composed of two RF pulses separated by a bipolar gradient [1-4]. The RF flip angles are α and $\alpha\exp(i\phi)$. If a velocity induced phase shift θ is obtained between the two pulses, the magnetization after the second pulse is tipped by $2\alpha\ \exp(i(\theta+\phi)/2)\cos((\theta-\phi)/2)$, assuming a small tip angle approximation. Let field-of-speed (FOS) be the velocity corresponding to a phase shift of 2π due to the effect of the bipolar gradient, then $\theta = 2\pi v/FOS$ for spin velocity v. If $\phi = 2\pi v_0/FOS$, for a given velocity v_0, then the magnetization is tipped as a function of velocity by the weighting function $\cos(\pi(v-v_0)/FOS)$. Bright bands are obtained for $v=v_0$ modulo FOS and dark bands for $v=v_0+FOS/2$ modulo FOS, which corresponds to a velocity tagging.

MATERIAL AND METHODS

Experiments were performed on a 1.5T GE Signa CV/i MR system with a maximum gradient strength of 33mT/m and a maximum slew rate of 120 T/m/s. The velocity-tagging pulse was incorporated into a 2D spoiled gradient echo sequence (Fig. 1). The first selective pulse was repeated with an inverted gradient polarity (and opposed frequency from the isocenter frequency). The 2nd pulse phase was varied to select v_0.

Flow phantom experiments: axial slice thickness=10 mm, flip angle=15˚, TR/TE=50/4.7 ms (TE given from 2nd RF pulse to echo), BW=±31.25 kHz, matrix size=256×128, FOV=6 cm×3 cm, FOS varied from 1 to 0.01 m/s (time between RF centers from 3 to 18 ms). Reception was realized with a 7.5-cm diameter surface coil. The flow phantom consisted of a flexible silicone tube (diameter 9.7 mm), straight for more than 70 cm and then curved with a curvature radius of 64 mm. A stationary flow rate was maintained to assure laminar flow (9.6 ml/s for the straight-tube experiment, 9.05 ml/s for the curved-tube experiment, measured by phase-contrast imaging). Measurements were performed upstream and downstream of the curvature. Simulations of the expected result on parabolic flow were also performed for comparison with the phantom images.

Human experiments: cine imaging of the common carotid artery, axial slice thickness=10mm, flip angle=15˚, TR/TE=33/5 ms BW=±31.25 kHz, matrix size 256×128, FOV=6 cm×3 cm, FOS=0.0915 m/s, cardiac triggered peripheral pulse gating, 18 phases and 2 views per segments.

RESULTS

For the parabolic flow pattern velocity rings were observed as expected (Fig. 2). The radii decreased as velocity increased towards the center of the tube. Rings were more concentrated and thinner towards the edge than at the center due to higher velocity gradient. By counting the rings, the maximum velocity was estimated to be 0.27 m/s (3 rings from 0), which is close to the theoretical and measured (with phase-contrast) value of 0.26 m/s. After the curvature the velocity distribution was asymmetrical with a maximum on the outer edge of the curvature and U-shape profile as correctly demonstrated by the velocity contour lines (Fig.3). Maximum velocity was estimated to be 0.18 m/s. In-plane velocity-tagged images indicated the presence of symmetrical eddies occurring in curvature. For the in vivo acquisitions (Fig. 4), various rings were seen depending on the cardiac phase and parabolic-like isovelocity rings were observed during diastole indicating a maximum velocity around 0.3 m/s.

DISCUSSION AND CONCLUSION

This velocity contour mapping technique provides a quick and easy way to study flow pattern and flow quantification by simply examining velocity contour lines. Compared to the traditional phase contrast method for mapping flow, this velocity contour mapping provides the benefits of fast (half scan time) and robust (higher SNR) flow examination. For accurate velocity quantification, a phase unwrapping algorithm may be employed to determine the velocity contour levels. While tagging precision can be affected by B_0 inhomogeneity (additive phase shift acquired between RF pulses), this was not observed in the presented experiments. The velocity tagging sequence can be further optimized by combining the slice encoding gradient with the bipolar gradient to reduce application time. By playing on the RF phases adequately, it is also possible not to tip static tissue. Such a technique could also be used to spoil the signal from moving spins if FOS is chosen small enough to produce signal cancellation within voxels.

In summary, a fast and easy velocity contour mapping technique is developed based on velocity tagging, and in vivo feasibility is shown. It may be a practical tool in addition to the traditional MR phase contrast methods.

REFERENCES 1. J.M. Pope, S. Yao, Magn Reson Imaging. 11, 585 (1993) **2.** D.G. Norris, C. Schwarzbauer, J Magn Reson. 137, 231 (1999) **3.** L. de Rochefort et al., Magn. Reson Med. 55, 171 (2006) **4.** H. Wen, Magn Reson Med. 46, 767 (2001)

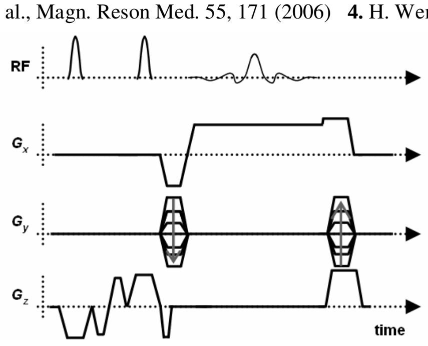

Fig 1. Through-plane velocity-tagging 2D sequence.

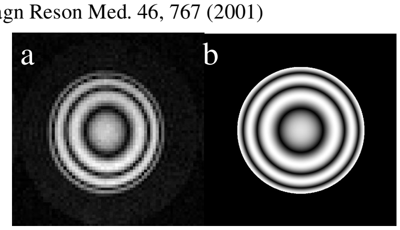

Fig 2. a) Velocity-tagged image obtained on a parabolic flow with FOS=0.0915 m/s, tagging applied normal to the plane, and b) simulated velocity-tagged image with the same parameters (flow rate, FOS)

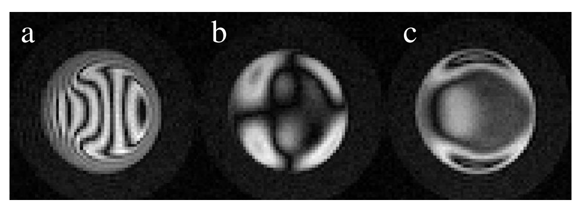

Fig 3. Velocity-tagged image obtained after the curvature with a) FOS=0.024 m/s for through-plane velocity, b) FOS=0.01 m/s for top-bottom velocity, and c) left-right velocity. Outer edge of the tube is located on the right of the images.

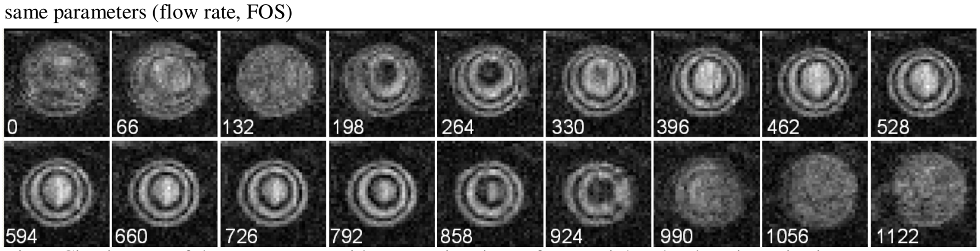

Fig 4. Cine images of the common carotid artery. Time in ms from peripheral pulse trigger is given.