Journal of Computer Assisted Tomography
18(3):469–475, May/June

Cine Phase-Contrast MR Flow Measurements: Improved Precision using an Automated Method of Vessel Detection

David J. Burkart, Joel P. Felmlee, C. Daniel Johnson, Ronald L. Wolf, Amy L. Weaver, and Richard L. Ehman

Objective: The purpose of this study was to construct a method of vessel edge detection that correctly identifies vessel pixels and to compare the interuser variability of cine phase contrast MR volumetric flow rates obtained with the conventional manual method and an automated method.

Materials and Methods: The automated method was developed based on a magnitude image threshold value and compared with the manual method in a flow phantom (three users) and in velocity images of the portal vein (five users). The threshold value determined from the magnitude image was applied to a region of interest surrounding the vessel of interest on the magnitude image to construct a vessel edge detection mask $m(x,y)$. The velocity images were then multiplied by the mask $m(x,y)$ and volumetric flow rates determined using the identified vessel pixels.

Results: In the flow phantom, flow measurements with the magnitude threshold method had significantly less interuser variability compared with the manual method ($p < 0.01$) and were within 10% (mean 6.0%) of the actual flow versus 35% (mean 18.6%) with the manual method. Regarding flow measurements in the portal vein of six volunteers, the magnitude threshold method was significantly more precise ($p < 0.01$) than the manual method with a mean standard deviation between the five users of 40.4 ± 12.9 ml/min (range 22–60 ml/min) and 110.4 ± 32.7 ml/min (range 70–155 ml/min), respectively.

Conclusion: The magnitude threshold method of vessel edge detection developed in this study yields flow measurements that are accurate in the model system and have significantly less interuser variability than the manual method. This method shows promise for improving the precision of cine phase contrast flow measurements.

Index Terms: Magnetic resonance imaging, cine—Magnetic resonance imaging, techniques—Flow, measurements.

Cine phase contrast MR techniques are being used with increased frequency to noninvasively analyze vascular systems. These techniques are promising for the anatomical display and quantitative hemodynamic evaluation of various arterial and venous structures (1–6). Before these techniques are widely applied clinically, the interuser variability of cine phase contrast flow measurements needs to be critically evaluated and the precision of the measurements optimized (7,8).

From the Departments of Diagnostic Radiology (D. J. Burkart, J. P. Felmlee, C. D. Johnson, R. L. Wolf, and R. L. Ehman) and Biostatistics (A. L. Weaver), Mayo Clinic, 200 1st St. SW, Rochester, MN 55905, U.S.A. Address correspondence and reprint requests to Dr. D. J. Burkart.

The cine phase contrast acquisition yields velocity images in which pixel value is proportional to flow velocity. In theory, the evaluation of the velocity images resulting from cine phase contrast imaging should not require exact vessel edge detection for accurate flow measurements (9). This is because the stationary background tissue should have a pixel value of zero, and differences in user-specified regions of interest (ROIs) surrounding the vessel in which flow is to be calculated should not alter the flow measurements. Accordingly some investigators have given limited attention, to date, to the variability of volumetric flow rates between users. Other groups have studied the difference in flow measurements in the great vessels of the thorax between two observers and have found mean ± SD differences of 8.02 ± 8.43% [% mean difference = (measurement 1 − measurement 2)/mean of both measurements] (2–4). Likewise, in the authors' experience, differences in individual flow measurements between users have varied by as much as 24% (D. J. Burkart et al., presented at Radiological Society of North America meeting, November 1992). In many potential clinical applications of cine phase contrast measurements, this level of variability is unacceptably high and needs to be improved.

The purpose of this study was to construct a method of vessel edge detection that correctly identifies vessel pixels and to compare the interuser variability of cine phase contrast volumetric flow rates obtained with the conventional manual method and the automated method. To our knowledge this is the first method of vessel edge detection designed to increase the precision of cine phase contrast flow measurements.

METHODS

Cine Phase Contrast Imaging

Phase contrast MR methods are based on the principle that the transverse magnetization of spins moving in the presence of a magnetic field gradient develops a velocity-proportional phase shift relative to static spins (10–12). Two acquisitions with different flow-encoding gradients along the same direction are interleaved in a cine sequence and subtracted to yield a velocity image in which the pixel value is proportional to the velocity of flow along the flow-encoded axis (13–16). Cardiac monitoring is used to control incrementing of the phase encoding during acquisition, and the data are retrospectively sorted into a user-specified number of cardiac phases (13,14). The technique typically produces 16 velocity image and magnitude image pairs spanning the cardiac cycle.

Manual Method of Volumetric Analysis

Currently the most commonly used method of volumetric analysis, the manual method, requires each user to evaluate the magnitude image and manually place an ROI surrounding the vessel in which flow is to be measured. The operator display console of the MR system is used to create standard circular, rectangular, and free-hand ROIs. The edge of the vessel is determined visually by each user, and the window and level settings are adjustable. The images are not magnified to select the ROI. The user-selected ROI is automatically transposed from the magnitude image to the corresponding velocity image. The area–velocity product of all the pixels within the ROI determines the volumetric flow rate. A volumetric flow rate is calculated for each of the 16 velocity images, and these rates are averaged to provide the mean flow during the cardiac cycle. Some investigators use a threshold value derived from the magnitude image to suppress background noise in the velocity image before manually selecting the ROI (13,17). No background noise suppression techniques are used in this study.

Since this method of volumetric analysis is a manual technique of vessel edge detection, the ROIs selected by each user may contain a variable number of nonvessel (background) pixels with nonzero pixel values (Fig. 1). Additionally, an ROI may also exclude true vessel pixels and underestimate the actual volumetric flow rate.

For the purposes of this study, each user was instructed to select an ROI that included the entire vessel of interest and excluded nonvessel background material (see Figs. 3a and 4a). Each user independently selected the optimum window and level setting and the shape of the ROI to accomplish this goal. The same ROI was applied to each of the velocity images since in this study the vessels were stationary and contained nonpulsatile flow.

Magnitude Threshold (Automated) Method of Volumetric Analysis

An automated method of vessel edge detection was developed based on two assumptions. In the magnitude image, vessel pixels have a higher intensity relative to nonvessel (background) pixels (Fig. 2). This is due to flow-related enhancement (11). Second, all pixels with intensity values in the mag-

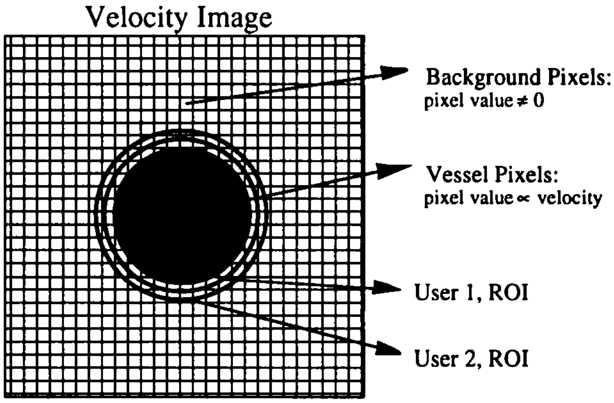


FIG. 1. Manual method of volumetric analysis. This diagram shows that in practice, background (nonvessel) pixels in the velocity image have a nonzero value and vessel pixels have a value proportional to flow velocity. Since flow measurements are determined from all the pixels contained within the region of interest (ROI), differences in the ROI selected by user 1 and user 2 will result in flow measurements that lack precision.

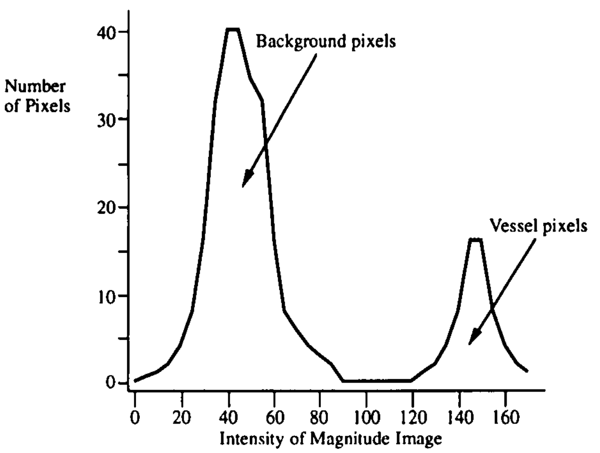


FIG. 2. Magnitude image histogram. This histogram represents a region of interest surrounding a vessel with its contiguous background. It shows that vessel pixels have a higher intensity than nonvessel (background) pixels in the magnitude image.

nitude image greater than some threshold value define vessel pixels.

The magnitude threshold method of volumetric analysis is performed using the operator display console of the MR system. In this method, two ROIs are placed on the magnitude image (Figs. 3b and 4b). The first ROI is generously placed surrounding the vessel in which flow is to be determined. The second ROI is placed in a nearby region of background. The size of these ROIs may be chosen to be the same or different. A magnitude threshold value is obtained from this background ROI.

The goal of the background magnitude threshold measurement is to reliably determine the peak intensity of the background tissue and not be affected by random intensity variations due to noise. We chose a simple average of the highest three intensities in the background ROI to minimize the effect of a single high intensity outlier point. Experiments with this simple model proved to yield reliable results. However, more robust statistical models and techniques may offer improved reliability and should be explored, especially to include the effects of disease on background tissue signal.

The threshold value determined from the magnitude image is applied to the first ROI on the magnitude image to construct a vessel edge detection mask $m(x,y)$. The mask $m(x,y)$ is created by assigning a value of 1 to pixels with intensities greater than the threshold intensity and a value of 0 to pixels with intensities less than the threshold intensity (Figs. 3c and 4c). Vessel pixels are defined as all pixels with intensities greater than the threshold value. The velocity images are then multiplied by the mask $m(x,y)$ and volumetric flow rates determined using the identified vessel pixels. The vessel detection mask $m(x,y)$ is displayed on both the magnitude and the velocity images to allow the user to assess the validity of the vessel edge determination (Figs. 3c, 4c, and 5). If a pixel with a value less than the threshold value is present within the lumen of the vessel, the mask $m(x,y)$ is defined as invalid.

EXPERIMENTS

All studies were performed with a 1.5 T imager (General Electric, Milwaukee, WI, U.S.A.) employing a gradient-recalled echo-based cine phase contrast pulse sequence that is commercially available (16).

Phantom Study

To investigate the accuracy of the magnitude threshold method for measuring volumetric flow

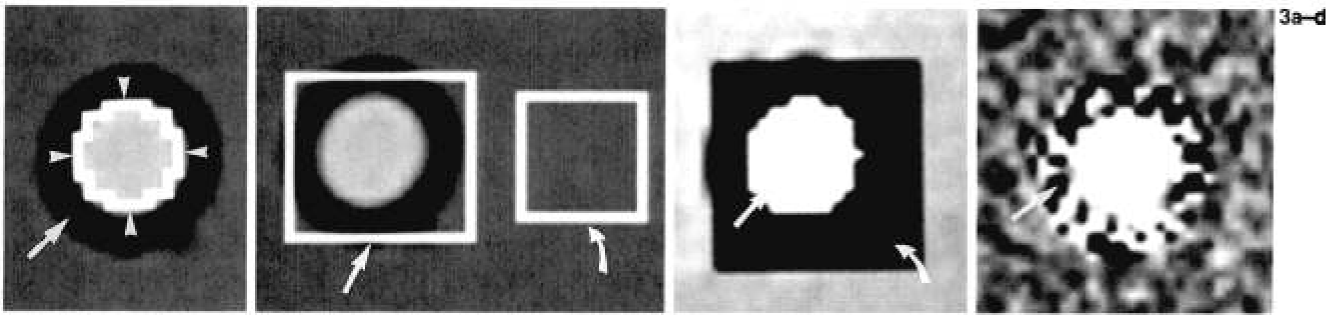


FIG. 3. Flow phantom images. **a:** This magnitude image of the 13 mm tube shows that there was no signal in the tubing wall (arrow). An example of the manual method is demonstrated by the white circular region of interest (ROI) placed around the flow within the tubing (arrowheads). **b:** The same magnitude image shows an example of the magnitude threshold method. Typical placement of the vessel ROI (straight arrow) and the background ROI (curved arrow) is demonstrated. **c:** The same magnitude image after the automatic construction of the mask $m(x,y)$ by the magnitude threshold method. The mask $m(x,y)$ displays pixels within the lumen of the tube (pixel value > threshold value) as high intensity (straight arrow) and pixels outside the lumen of the tube (pixel value < threshold value) as low (black) signal intensity (curved arrow). Only pixels identified by the mask $m(x,y)$ as vessels pixels are used in flow calculations. **d:** The corresponding velocity image of the 13 mm tube shows nonzero pixel values within the tubing wall (arrow) and static background. These nonzero values may cause flow measurements between users with the manual method to be variable.

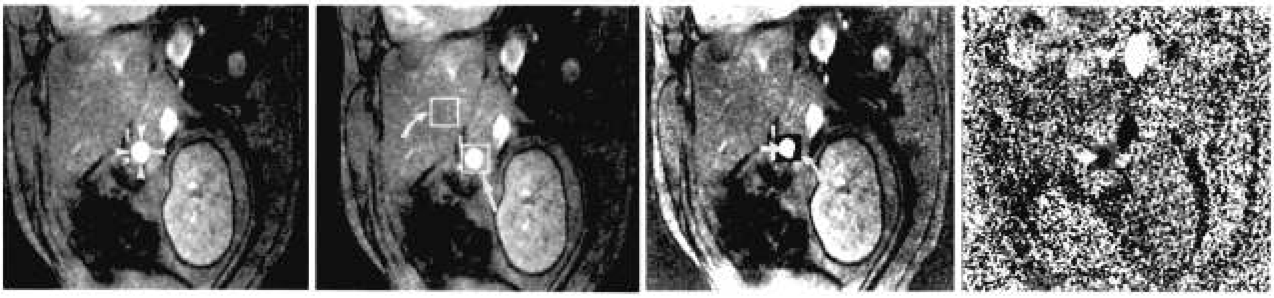
4a–d


FIG. 4. Portal vein images. **a:** This magnitude image of the portal vein shows a typical example of the manual method. A white circular region of interest (ROI) is placed around the portal vein (arrowheads) to include the entire vessel and exclude adjacent static background. **b:** The same magnitude image shows an example of the magnitude threshold method. Typical placement of the vessel ROI (straight arrow) and the background ROI (curved arrow) is demonstrated. The vessel ROI is generously placed to include the entire portal vein. The background ROI is placed in a homogeneous region of the liver. **c:** The same magnitude image after the automatic construction of the mask $m(x,y)$ by the magnitude threshold method shows pixels within the lumen of the portal vein as high intensity (straight arrow) and pixels outside the lumen of the portal vein as low (black) signal intensity (curved arrow). This mask is multiplied by the corresponding velocity image to determine which pixels will be included in the flow calculation. **d:** The corresponding velocity image after the automatic construction of the mask $m(x,y)$ by the magnitude threshold method shows pixels within the lumen of the portal vein as low intensity (straight arrow) and pixels outside the lumen of the portal vein as intermediate signal intensity (curved arrow). There are nonzero pixel values within static nonvessel tissues. These nonzero values may be due to susceptibility effects, microcirculation effects, and engineering reliability issues.

rates and to compare the interuser variability of flow measurements with the magnitude threshold method and the manual method of volumetric analysis, a flow phantom was constructed. Two flexible tubes (inner diameters 13 and 6 mm) were immersed in a fluid with a T1 approximating that of liver tissue. A second solution doped with gadopentetate dimeglumine and $MnCl_2$ to yield a T1/T2 of 847/85 ms was pumped through the tubing in a nonpulsatile fashion at independently measured flow rates ranging from 400 to 2,300 ml/min, determined by timed filling of a graduated cylinder.

Cine phase contrast MR images perpendicular to the flexible tubing were acquired with the following parameters: TR, 33 ms; TE, 7.8 ms; flip angle, 20°; no. of excitations, 2; phase-encoding views, 256; section thickness, 5 mm; field of view, 28 cm. Velocity encoding was accomplished along the section selection gradient that coincided with the true direction of flow. Velocity encoding to cause a π phase shift was chosen as twice the mean velocity in the tube in which flow was to be determined. The mean velocity was determined as the actual volumetric flow rate divided by the area of the lumen of the tubing. Separate acquisitions were obtained for each tube. A simulated electrocardiogram at 100 beats/min was used to trigger the sequence. Sixteen phases of the cardiac cycle were reconstructed.

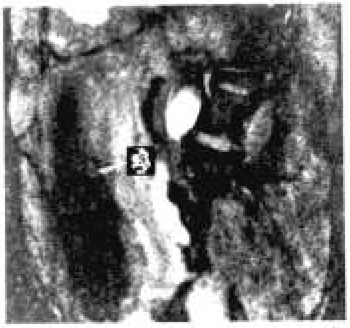


FIG. 5. Graphical feedback mechanism. This magnitude image shows inappropriate vessel edge detection of the portal vein by the magnitude threshold method. Mask $m(x,y)$ (arrow) displays the contour of the portal vein as irregular and includes low (black) intensity nonvessel pixels within the lumen of the portal vein. Flow measurements with pixels identified by this mask $m(x,y)$ will be invalid. This mask $m(x,y)$ was the result of incorrectly placing the background region of interest (ROI) required by the magnitude threshold method of volumetric analysis. Subsequently, an appropriate background ROI was selected and appropriate vessel edge detection was achieved.

Three users applied each method of volumetric analysis to the resulting sets of images. The users were blinded to the actual volumetric flow rates. For the magnitude threshold method, the users were instructed to select the background ROI in the static fluid surrounding the flexible tubing (Fig. 3c). For the manual method of volumetric analysis, each user was instructed to select an ROI including the moving fluid within the tube and excluding the tubing wall and static background fluid (Fig. 3b).

Based on the measurements obtained by the three users, the interuser variability of each method was estimated by determining the coefficient of variation [(SD/mean) × 100] at each volumetric flow rate. The accuracy of each method was assessed based on calculating the percentage error for each measurement [(measured volumetric flow rate − actual volumetric flow rate) × 100/actual volumetric flow rate]. The absolute value of each percentage error was also obtained and averaged across the three users to obtain an estimate of the mean absolute percentage error at each volumetric flow rate. The results were summarized across the volumetric flow rate levels by presenting the mean and standard deviation separately for each method of flow analysis and tube size. Comparisons between the two methods were made using a paired t test, whereas comparisons between the 6 and 13 mm results within a given method were made using a two sample t test. All tests were two sided and p values of <0.05 were considered statistically significant.

In Vivo Study

To compare the interuser variability of the magnitude threshold method versus the manual method of volumetric analysis, cine phase contrast images were obtained in the portal vein of six healthy volunteers (mean ± SD age, 32.8 ± 5.9 years) after fasting 6 h. These images were acquired perpendicular to the main portal vein with the following parameters: TR, 33 ms; TE, 7.8 ms; flip angle, 30°; velocity encoding, 40 cm/s; no. of excitations, 2; phase-encoding views, 256; section thickness, 5 mm; field of view, 28 cm. Velocity encoding was accomplished along the section selection gradient that coincided with the true direction of flow. These images were acquired with respiratory-ordered phase encoding and plethysmographic gating, with reconstruction of 16 phases of the cardiac cycle.

Five independent users evaluated the images obtained in the volunteers to determine volumetric flow rate with both the magnitude threshold method and the manual method of volumetric analysis. The users were instructed to select the background ROI (required in the magnitude threshold method) in the nearby liver parenchyma that was of homogeneous signal intensity and unoccupied by vessels (Fig. 4b). Each user was advised concerning the manual method to select an ROI to include the entire portal vein and exclude the background (nonvessel) material (Fig. 4a). All users had previous experience with the manual method of volumetric analysis.

To evaluate the interuser variability of each method, the coefficient of variation was estimated for each method based on the six measurements. Comparison between the two methods of the average difference in the coefficients of variation was made using a paired t test. Agreement between the two methods was assessed by obtaining the paired differences of the flow measurements between the two methods and calculating the bias, as estimated by the mean difference. The percentage error with each method was not determined because a standard of reference for portal venous flow is not available.

RESULTS

Phantom Study

A summary of the performance of the magnitude threshold method and the manual method of volumetric analysis in the flow phantom is shown in Table 1. The magnitude threshold method had significantly less interuser variability compared with the manual method in the 13 mm tube ($p = 0.0005$). Similarly, in the 6 mm tube the flow measurements obtained with the magnitude threshold method were more reliable than with the manual method ($p = 0.01$). Regarding the manual method, the interuser variability was greater in the 13 than 6 mm tube ($p = 0.002$).

In the 13 mm tube flow measurements with the magnitude threshold method were more accurate than measurements employing the manual method ($p = 0.005$). The mean percentage error was also less in the 6 mm tube for the magnitude threshold method than for the manual method ($p = 0.05$). The mean differences between the percentage error in flow measurements with the magnitude threshold method and the conventional method in the 13 and the 6 mm tubes were 11.92 and 13.26%, respectively. Differences in accuracy of flow measurements with the manual method of volumetric analysis in the 13 and 6 mm tubes were not significant ($p = 0.46$).

In Vivo Study

Table 1 summarizes the interuser variability observed with the magnitude threshold method and the manual method of volumetric analysis in flow measurements of the portal vein. There was significantly less variability between users of flow measurements obtained with the magnitude threshold method compared to the manual method ($p = 0.009$). The mean standard deviation of flow measurements between the five users with the magnitude threshold method was 40.4 ± 12.9 ml/min (range 22–60 ml/min), and it was 110.4 ± 32.7 ml/min (range 70–155 ml/min) for the manual method. Figure 6 shows the mean ± SD of flow measurements determined by the five users with both methods of volumetric analysis for each case.

The bias between flow measurements acquired with the magnitude threshold method and the man-

TABLE 1. *Interuser variability and precision with each method of volumetric analysis*

	13 mm tube		6 mm tube		Portal vein	
	MTM	MM	MTM	MM	MTM	MM
Variability						
Mean (CV)	0.18	7.84	0.37	3.81	3.12	8.42
SD (CV)	0.12	2.34	0.18	1.83	0.91	2.45
p value	0.0005		0.01		0.009	
Accuracy						
Mean (| % error |)	5.89	17.81	6.22	19.48	—	—
SD (| % error |)	0.65	6.38	3.08	8.85	—	—
p value	0.005		0.05		—	

MTM, magnitude threshold method of vessel edge detection; MM, manual method of vessel edge detection; CV, coefficient of variation.

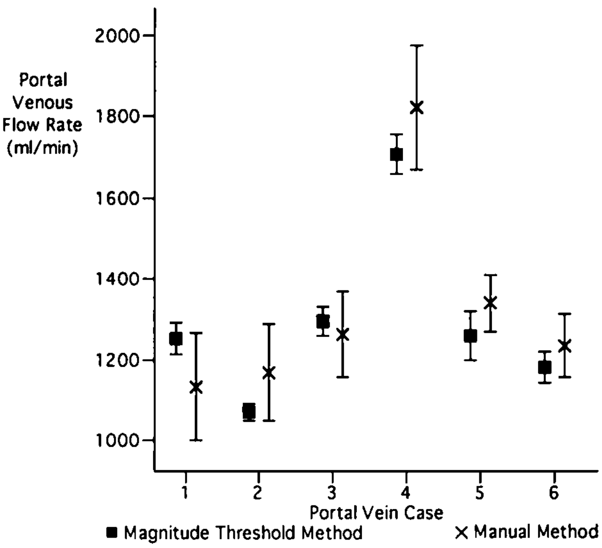


FIG. 6. Precision of portal venous flow measurements. This graph displays the mean ± SD (five users) of flow measurements in the portal vein with the magnitude threshold method and the manual method of volumetric analysis. It shows that flow measurements with the magnitude threshold method were more precise than those with the manual method.

ual method of volumetric analysis was −32.6 ml/min. This difference in flow measurements between methods was not statistically significant ($p = 0.42$). The percentage error with each method was not determined because a standard of reference for portal venous flow is not available.

DISCUSSION

Current literature suggests that it is important in clinical research to assess the interuser variability of MR techniques (7,8). The results of this study indicate that cine phase contrast flow measurements with the magnitude threshold method of vessel edge detection developed in this study have significantly less interuser variability than those obtained with the manual method of volumetric analysis. This improvement in the precision of flow measurements between users was observed in a flow phantom as well as in vivo. In the model system, flow measurements with the magnitude threshold method were more accurate than measurements with the manual method. A primary reason for the lack of precision between users with the manual method was that nonvessel background pixels in the velocity images were nonzero with a distribution that was not completely random.

Given that cine phase contrast techniques are being used with increased frequency as a noninvasive method for quantitative analysis of the vascular system, the precision of these flow measurements must be optimized. Although it has been hypothesized that phase contrast flow measurements, especially in vivo, are tolerant to interuser variability, the findings of this study have demonstrated that significantly high interuser variability is observed with the manual method of volumetric analysis in routine practice. Likewise, other investigators have observed mean differences in flow measurements with the manual method between only two users of 8.02 ± 8.43% (2–4). This amount of variability with cine phase contrast flow measurements may hinder clinical applications of this technique. Furthermore, all the users in this study had previous experience with the manual method of volumetric analysis. In a novice group of users, the variability between users with the manual method would likely be more pronounced.

The magnitude threshold method of volumetric analysis shows promise for improving the precision of cine phase contrast flow measurements. This method provided flow measurements with low interuser variability in the flow phantom and the portal vein, even though only a minority of the users had experience with the method. This suggests the feasibility of the magnitude threshold method of volumetric analysis.

The variability of cine phase contrast flow measurements between users with the manual method is due largely to the nonzero values of the nonvessel background pixels in the velocity image (Figs. 3d and 4d). Our flow measurements have shown that the statistically varying noise of the background is not totally random, so its mean value is not zero, thereby affecting the flow measurements. The nonzero background may be secondary to susceptibility effects, microcirculation effects, and engineering reliability issues. Susceptibility effects cause local magnetic field differences within the background tissues that result in varying phase evolution between the velocity-encoded and velocity-compensated acquisitions. As a result there is a net phase change in the background after subtraction of the velocity-encoded and velocity-compensated images. Microcirculation in the background tissues may also produce net phase changes. Finally, if there is radiofrequency phase instability or problems with gradient repeatability, echo centering, or echo sampling, the background tissues will have nonzero values in the velocity image. Cine phase contrast flow measurements with the manual method will also be variable between users if true vessels pixels are inadvertently excluded during placement of the vessel ROIs.

The magnitude threshold method employs vessel edge detection based on the intensities of the magnitude image to evaluate the velocity image. The mask $m(x,y)$ derived from the magnitude threshold demonstrates which pixels are used in determining flow measurements (Figs. 3c and 4c). The construction of this mask will not be valid if the background

ROI is incorrectly placed or if flow voids within the vessel of interest occur in the magnitude images. Flow void due to intravoxel dephasing is a difficult problem in flow imaging especially in the vicinity of a stenosis. This loss of signal will affect the mask determination and alter (generally decrease) the flow measurement. This potential limitation reinforces the role of the user (radiologist) in this method. All images must be inspected before this technique is applied. Furthermore, there is a need for a graphical feedback mechanism to allow the user to interpret the selection of the vessel pixels. The display of the mask $m(x,y)$ allows the user to evaluate if appropriate segmentation of the vessel is achieved (Figs. 4d and 5). This feedback strengthens the reliability of the magnitude threshold method.

The magnitude threshold method was also found to provide significantly more accurate flow measurements in the model system than the manual method. However, conclusions regarding the accuracy of cine phase contrast flow measurements with the manual method must be guarded, since it is likely that the phantom design with no signal in the thick wall of the tubing contributed to the inaccuracy of the flow measurements with this method. The magnitude signal of the tubing wall is essentially zero, while the phase or rather pixel value in the velocity image is nonzero in this same region. This is analogous to imaging air. Accordingly the users were instructed concerning the manual method to select an ROI that included the entire vessel of interest and excluded the tubing wall and static background material. Undoubtedly many of the ROIs included a small number of pixels containing the tubing wall, which may have adversely affected the accuracy of the flow measurements more than an in vivo vessel wall. This again emphasizes the necessity of correct vessel edge detection to obtain accurate flow measurements and the difficulties associated with manually placing an ROI surrounding the vessel in which flow is to be measured. The fact that interuser variability with the manual method of volumetric analysis was similar in the 13 mm tube and the portal vein suggests that adverse effects of nonvessel pixels with the manual method on the accuracy and precision of in vivo flow measurements cannot be disregarded.

In conclusion, the magnitude threshold method of vessel edge detection developed in this study yields flow measurements that are accurate in the model system and have significantly less interuser variability than the manual method of volumetric analysis. This method shows potential for improving the precision of cine phase contrast flow measurements.

REFERENCES

1. Mohiaddin RH, Wann SL, Underwood R, Firmin DN, Rees S, Longmore DB. Vena caval flow: assessment with cine MR velocity mapping. *Radiology* 1990;177:537–41.
2. Kondo C, Caputo GR, Semelka R, Foster E, Shimakawa A, Higgins CB. Right and left ventricular stroke volume measurements with velocity-encoded cine MR imaging: in vitro and in vivo validation. *AJR* 1991;157:9–16.
3. Caputo GR, Kondo C, Masui T, Geraci SJ, Foster E, O'Sullivan MM, Higgins CB. Right and left lung perfusion: in vitro and in vivo validation with oblique-angle velocity-encoded cine MR imaging. *Radiology* 1991;180:693–8.
4. Dulce MC, Mostbeck GH, O'Sullivan MM, Cheitlin V, Caputo GR, Higgins CB. Severity of aortic regurgitation: interstudy reproducibility of measurements with velocity-encoded cine MR imaging. *Radiology* 1992;185:235–40.
5. Van Rossum AC, Visser FC, Hofman MBM, Galjee MA, Westerhof N, Valk J. Global left ventricular perfusion: noninvasive measurement with cine MR imaging and phase velocity mapping of coronary venous outflow. *Radiology* 1992; 182:685–91.
6. Burkart DJ, Johnson CD, Morton MJ, Wolf RL, Ehman RL. Volumetric flow rates in the portal venous system: measurements with cine phase-contrast MR imaging. *AJR* 1993;160: 1113–8.
7. Beam CA, Sostman HD, Zheng JY. Status of clinical MR evaluations 1985–1988: Baseline and design for further assessments. *Radiology* 1991;180:265–70.
8. Markus JB, Somers S, Franic SE, Moola C, Stevenson GW. Interobserver variation in the interpretation of abdominal radiographs. *Radiology* 1989;171:69–71.
9. Wolf RL, Ehman RL. Determinants of accuracy and precision in MR flow volumetry. *Magn Res Med* (in press).
10. Hahn EL. Detection of sea-water motion by nuclear precession. *J Geophys Res* 1960;65:776–77.
11. Axel L. Blood flow effects in magnetic resonance imaging. *AJR* 1984;143:1157–66.
12. Moran PR. A flow zeugmatographic interface for NMR imaging in humans. *Magn Res Imag* 1982;1:197–203.
13. Nayler GL, Firmin DN, Longmore DB. Blood flow imaging by cine magnetic resonance. *J Comput Assist Tomogr* 1986; 10:715–22.
14. Glover GH, Pelc NJ. A rapid cine MRI technique. In: Kressel HY, ed. *Magnetic resonance annual 1988*. New York: Raven Press, 1988:299–333.
15. Pelc NJ, Shimakawa A, Glover GH. Phase contrast cine MRI. *Proc SMRM* 1989;101.
16. Pelc NJ, Herfkens RJ, Shimakawa A, Enzmann DR. Phase contrast cine magnetic resonance imaging. *Magn Res Q* 1991;7:229–54.
17. Firmin DN, Nayler GL, Klipstein RH, Underwood SR, Rees RSO, Longmore DB. In vivo validation of MR velocity imaging. *J Comput Assist Tomogr* 1987;11:751–6.