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Measurement of flow velocity by magnetic resonance imaging using 2D phase contrast technique: estimation of oblique flow

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ABSTRACT

This study analyzed the effects of the obliqueness of flow on the accuracy of measuring flow velocity by means of the 2D phase contrast MR technique. A constant flow phantom consisting of a pump, a polyethylene tube and a flow meter was assembled. A commercial 1.5 Tesla MR imager was used to perform flow velocity measurements. The phase contrast technique was used to estimate the flow velocity of saline through the phantom. The effects of changes in obliqueness of flow to the plane of imaging were studied. The obliqueness had a significant effect that was more pronounced with increasing section thickness. An increase in the obliqueness angle caused underestimation of the average and maximum velocities. The obliqueness was found to be an important parameter affecting the estimation with the 2D phase contrast MR technique.

The application of MR to flow velocity measurement was first described by Singer[1)]. Various pulse sequences have been proposed to evaluate the flow velocity by MR. The two main classes of techniques are the time-of-flight method and the phase contrast method. The time-of-flight technique derives its contrast from the flow-related enhancement of inflowing blood[1～10)]. The phase contrast method, on the other hand, is based on applying balanced gradient pulses, and derives its contrast by detecting spin phase differences as blood moves across a magnetic field gradient[2)]. The phase contrast technique is not only a simple subtraction of phase images on a pixel by pixel basis but a complex difference of phase obtained with a known different flow encoding gradient. Phase images are collected using balanced gradient pulses to produce a phase shift for moving spins. The difference in phase is directly proportional to the flow velocity of the spins within the corresponding voxels[11～15)].

The phase contrast technique is generally better than the time-of-flight technique for quantitative measurement of flow velocity in cardiovascular an cerebrospinal abnormalities[16～19)]. Advantages of the phase contrast method include bi-directional flow sensitivity, ability to obtain both a conventional magnitude image an a velocity image from the same data set[10)] and the relative simplicity in interpretation, processing, and display[15)].

Cardiac gating can be used in conjunction with phase contrast to provide flow velocity encoded phase images at equally spaced temporal intervals during the cardiac cycle. Interpretation of flow velocity values during the cycle can provide instantaneous average and maximum flow velocity information[20)].

The purpose of this study is to evaluate the effects of obliqueness of the flow on the accuracy

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of the MR flow velocity measurement.

MATERIALS AND METHODS

Flow Phantom;

We studied flow velocity of saline in a polyethylene tube with an inside diameter of 6.4 mm; inside area of 0.32 cm^2. The tube was connected in a loop with a fluid supply reservoir. Two sections of the tubing, one with outflow fluid and the other with return flow, were embedded in the center of a 10% agar-gel container (25 cm length, 17 cm width, 10 cm height). The phantom was placed in the MR imager with the direction of flow parallel to the static magnetic field vector (i.c.; along the z axis).

The constant flow was driven by a pump (BIO-CONSOLE$_{T.M.}$; VIO-MEDICUS, INC., Minnetonka, MN, USA). The pump was placed outside the magnet room to minimize an interaction between pump and magnet, and to reduce radio frequency interference[21]. The fluid used was saline with a T1 relaxation time of approximately 4065 msec.

Imaging Setup;

The flow phantom was placed ina a 1.5 Tesla clinical MR imager, GE Signa (General Electric MEdical Systems, Milwaukee, WI, USA). The two-dimensional (2D) phase contrast technique using balanced gradient pulses was used to measure the flow velocity of the fluid. Measurements were performed using a circular surface coil with a diameter of 12 cm. A field of view of 16 cm was used, and the slice thickness was 10 mm. The imaging parameters used were repetition time (TR) = 50 msec, echo time (TE) = 12.1 msec, the flip angle was 300, the matrix size was 256 × 256, and number of acquisition was one. We confirmed the accuracy of 2D phase contrast MR flow velocity measurement by using a flow meter, and then began the study to test the impact of changing the obliqueness of flow.

The effect of oblique flow was studied in two different slice thicknesses (3 mm and 10 mm) by changing the tube angle to the imaging plane in steps of 10^0 up to 80^0. We made 28.6 cm/sec average flow velocity by a pump and 70.0 cm/sec VENC (velocity encoding) value were used. We evaluated the maximum flow velocity in this study due to difficulty in creation of an elliptic ROI. Maximum flow velocity was obtained by determining the pixel with the maximum flow velocity within the tube area. We measured the maximum flow velocity components along the X, Y, and Z axes, and calculated the MR measured maximum flow velocity ($V_{(MR)}$) through the following equation (1),

$$V_{(MR)} = \sqrt{V_{X(MR)}{}^2 + V_{Y(MR)}{}^2 + V_{Z(MR)}{}^2} \qquad (1)$$

where $V_{X(MR)}$, $V_{Y(MR)}$ and $V_{Z(MR)}$ are the MR measured maximum flow velocity components in the X, Y, and Z directions, respectively. When the flow direction was angled from the Z axis (slice selection direction, which is normal to the slice plane) towards the X axis (frequency encoding direction) in the X-Z plane, the relationship between the true maximum flow velocities and the true maximum flow velocity components along the X axis and the Z axis are

$$V_X = V \sin q, \qquad (2)$$
$$V_Z = V \cos q, \qquad (3)$$

where q is angle between the flow direction and the slice selection direction, and V_X, V_Z, and V are the true maximum flow velocity components along the X axis, the Z axis, and true maximum flow velocity, respectively.

We compared the true maximum flow velocity with the MR measured maximum flow velocity from the MR measured maximum flow velocity vomponents according to equation (1), and we also compared the MR measured maximum flow velocity components along the X and Z axes with the calculated true maximum flow velocity components according to equation (2) and (3). As a function of obliqueness, the "true" maximum flow velocity was defined as the MR measured maximum flow velocity at q = 0.

RESULTS

Obliqueness of the flow to the imaging plane had a significant effect on the accuracy of flow velocity measurement with 2D phase contrast. As shown in Fig-1a and 1b, an increase in the oblique angle caused underestimation of the maximum flow velocities. The obliqueness effect was more pronounced when the slice thickness was increased from 3 mm to 10 mm. As the angle of obliqueness increased from 0^0 to 80^0, measurements of maximum flow velocities decreased from 54.4 cm/sec to 34.9 cm/sec for 10 mm slices, and from 54.7 cm/sec to 42.1 cm/sec for 3 mm slices.

Errors in flow velocity measurements were primarily observed along the frequency encoding direction (X axis) (Fig. 1a, b). MR measurements of the maximum flow velocity along the slice selection direction (Z axis) correlated well with the maximum flow velocity component along the Z axis calculated by equation (3). The agreement between the MR measured maximum flow velocities and true maximum flow velocities was good up to oblique angles of 60^0 for 3 mm slice thickness and 40^0 for 10 mm slice thickness. Although virtually zero flow velocity was detected in the Y direction since the tube was rotated in the X-Z plane about the Y axis, actually the MR measured maximum flow velocity in the Y direction became around 2 cm/sec because of noise of a background.

DISCUSSION

Phase contrast flow velocity measurement is limited, because the spine phase is cyclic and there is a natural limit of 2 radians to the maximum range of phase that can be measured[22]. The range of flow velocities that can be measured by the phase contrast technique covers the complete physiological range from approximately 0.1 mm/sec[23] to 500 cm/sec[24].

The MR phase contrast technique is sensitive to flow both within and perpendicular to the imaging plane[25], but flow-induced phase shift occurs only is the direction of the applied balanced gradient pulse[20]. Although it is possible to measure the flow velocity components in all three orthogonal directions (X, Y and Z axes), flow velocity in the oblique directions cannot be measured directly by our MR equipment. So we had to use equation (1) to calculate the velocities of flow oblique to the imaging plane.

We compared the true maximum flow velocity and the true maximum flow velocity components along X (frequency encoding direction) and Z (slice selection direction) axes with the calculated MR measured maximum flow velocity and the MR measured maximum flow velocity components along X and Z axes. As a function of the angle of the "true" maximum flow velocity was defined as the MR measured maximum flow velocity components were calculated by equations (2) and (3). True and MR measured maximum flow velocity component along the Z axis (V_Z and $V_{Z(MR)}$) were the same up to 80^0 obliqueness, and the true and MR measured maximum velocities also correlated well, regardless of the slice thickness, however, true maximum flow velocity components along the X axis (V_X) was inconsistent with MR measurement ($V_{X(MR)}$) at larger oblique angles (exceeding 40^0). The error in MR measured maximum flow velocity component along the X axis increased when the slice thickness increased.

The error is thought to be due to two effects. First, when a 10 mm slice thickness was used, there was a partial volume effect (i,e., phase shifts averaging of moving spins and static spins). When the phase average over a pixel includes a static phase shift due to spins outside the tube, the MR measured flow velocity will be underestimated. The percentage of static spins increases as the angle of obliqueness increases. In slices thinner than the tube diameter, however, there is less partial volume effect (Fig. 2a, b, c). The second reason for underestimation of the flow velocity is that the flow profile is not constant[26]. The flow near the tube wall is slower than at the center of the tube. The measured flow velocity is calculated by an average of various phase shifts due to different flow values dependent on the flow profile. While this error in measurement is not high even at a very high obliqueness angle (70^0), the error was nevertheless present and may account for the error observed in the 3 mm slices (Fig. 3a, b). We have no explanation why only $V_{X(MR)}$ had errors and $V_{Z(MR)}$ correlated with V_Z on both experiments.

In velocity measurements of oblique flow, the angle of obliqueness, the diameters of vessels, and the slice thickness all affect the accuracy of the flow velocity measurement. It is evident that slice thickness should be kept as small as possible to minimize the errors of oblique flow velocity measurement. This is particularly important in tortuous and/or winding vessels where larger obliquity can occur.

It is also possible to reduce the effect of obliqueness in MR imaging by carefully select-

Fig. 1a

Oblique Flow (10 mm Slices)

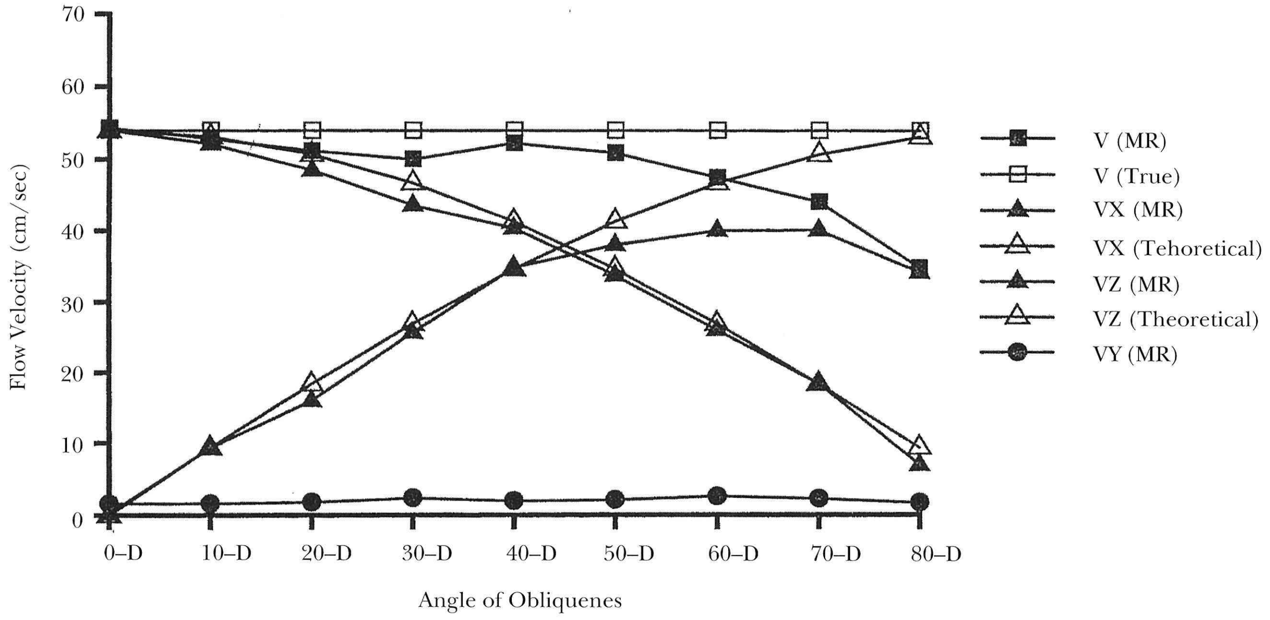

Fig. 1b

Oblique Flow (3 mm Slices)

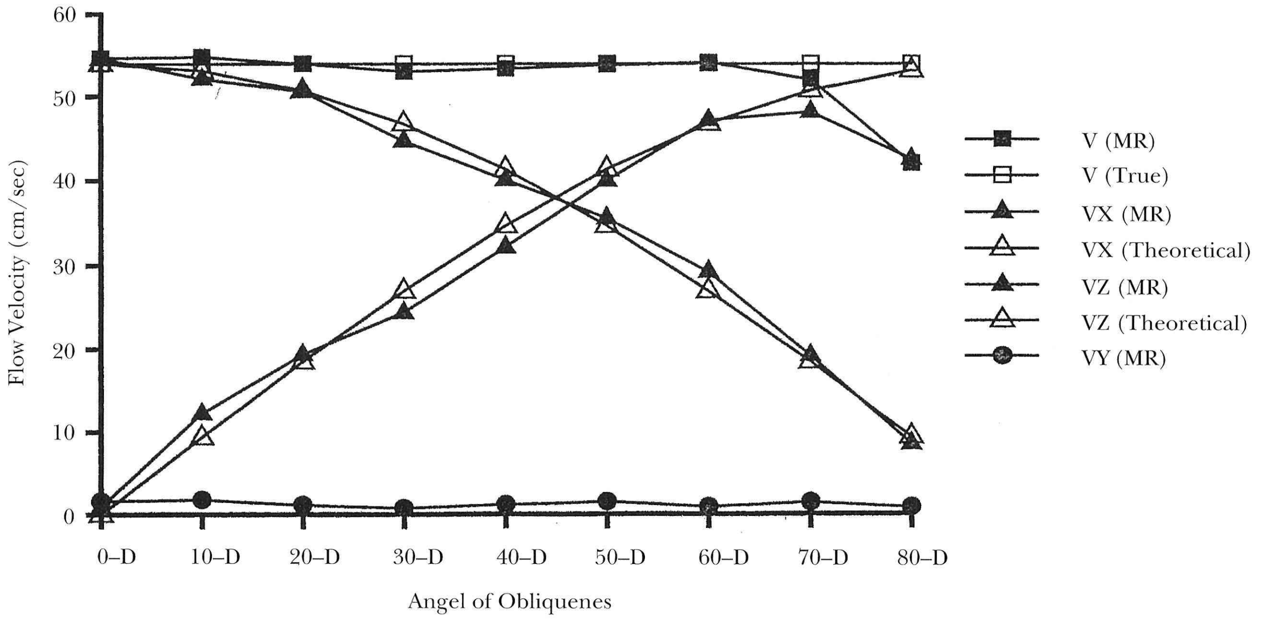

Fig. 1 True maximum velocity along the z axis measured through MR correlates well with actual V_Z measurements up to an 80^0 angle of obliqueness, regardless of slice selection. (a) Results of maximum velocity measurement using 10 mm slice thickness. Errors in measuring V_X are apparent where angle of obliqueness reached 50^0. (b) results of maximum velocity measurement using 3 mm slice thickness, error in measuring V_X are apparent when the angle of obliqueness reached 70^0.

Fig. 2a

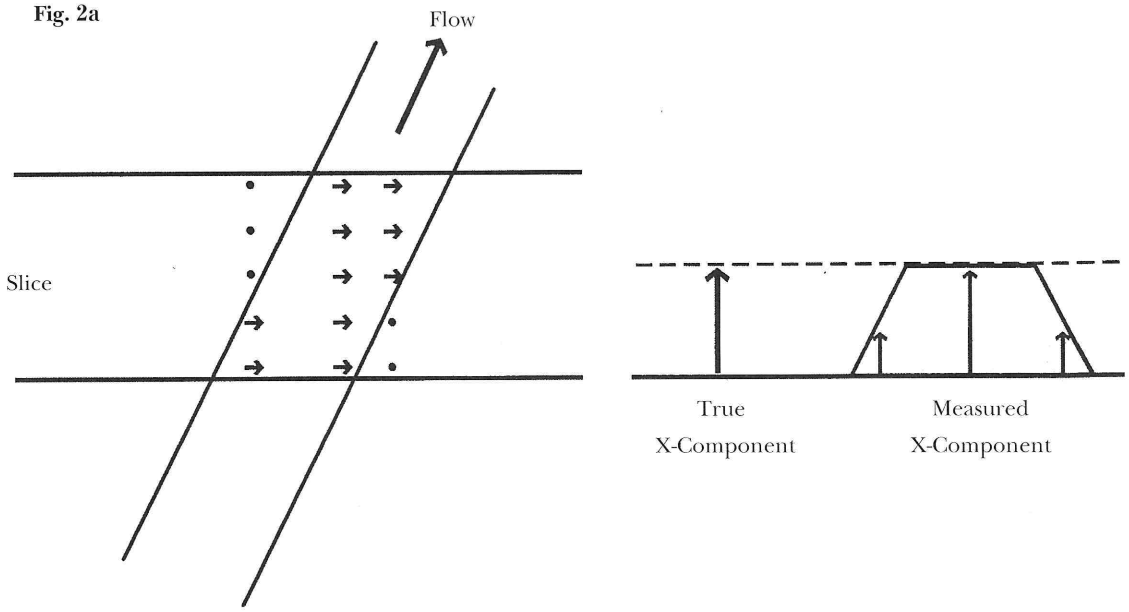

Fig. 2b

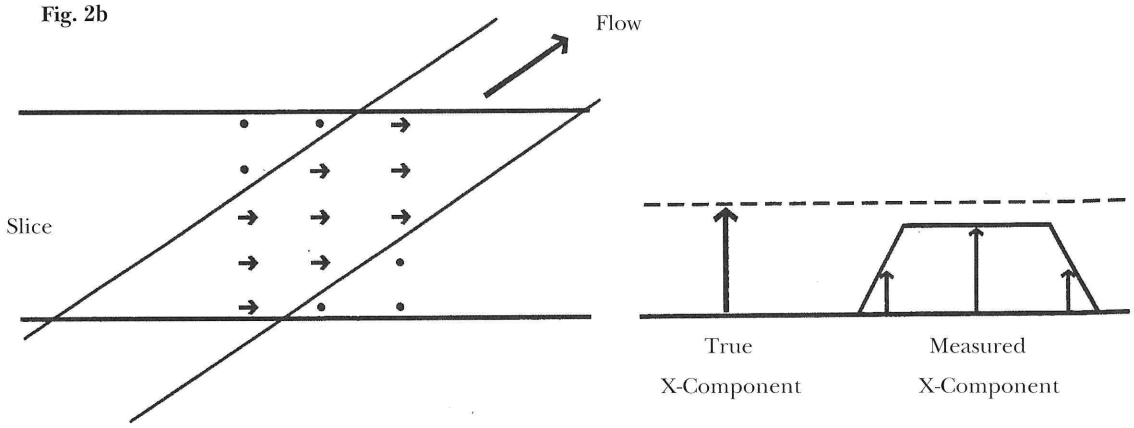

Fig. 2c

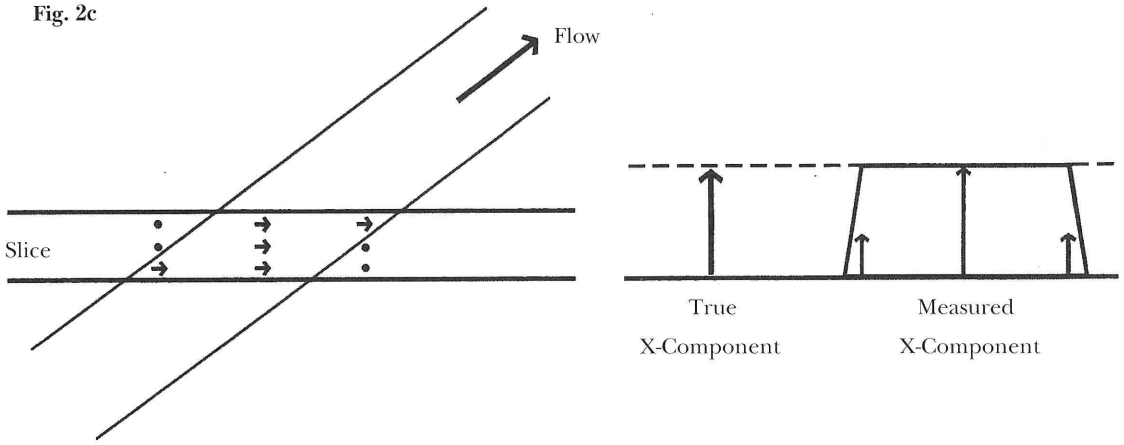

Fig. 2 (a) When a 10 mm slice thickness is used for the oblique flow measurement, there is partial volume effect of static spins (dots) and X component of moving spins (small arrows). When the phase average over a pixel includes a static phase shift due to spins outside the tube, the measured maximum flow velocity will be understimated. (b) The percentage of static spins increases as the angle of obliqueness increases. In situation (b), Every pixel includes the phase of static spins, so measured maximum X component velocity by MR is not compatible with the true maximum X component velocities. (c) Thinner slice thickness has less error. This explanation accounts for the error in measurement of the Z component velocity.

Fig. 3a

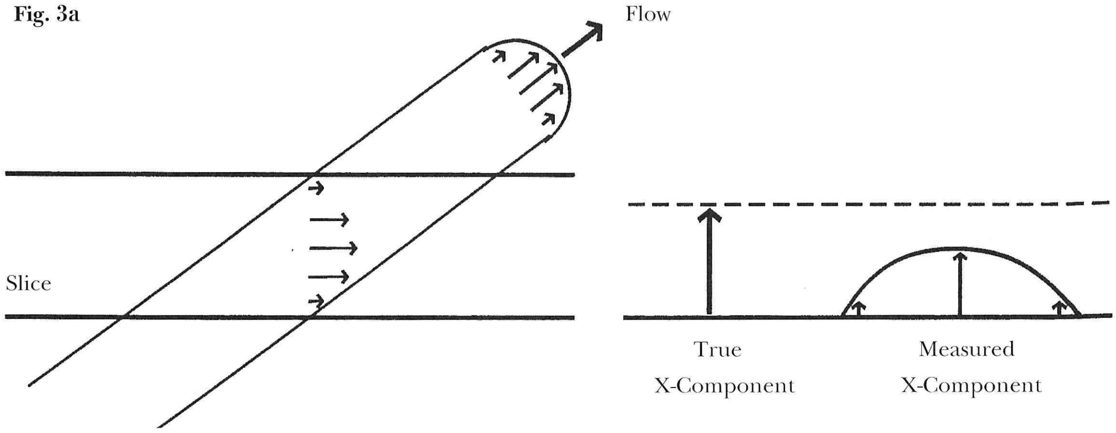

Fig. 3b

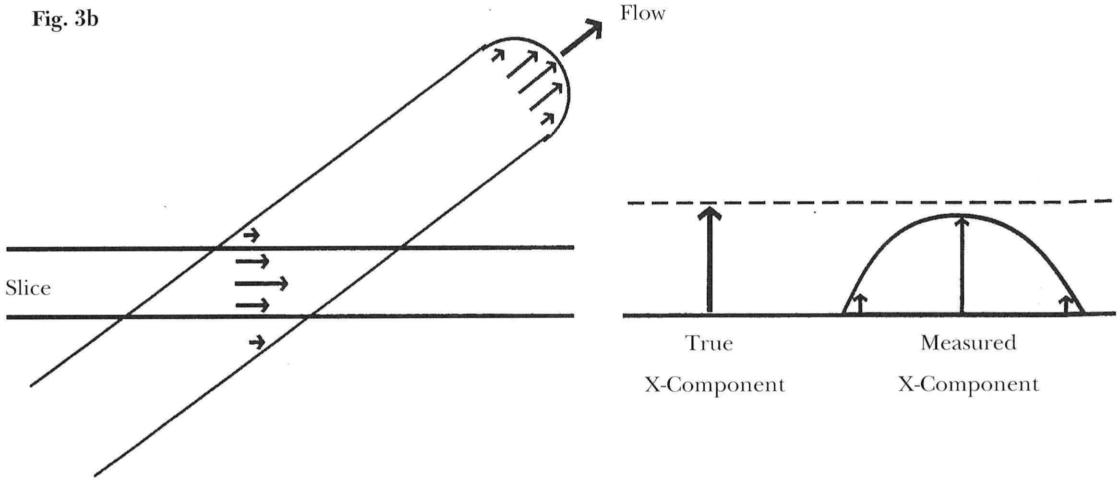

Fig. 3 (a) The flow near the tube wall is slower than at the center of the tube because the flow profile is not constant. The measured velocity is calculated by an average of various phase shifts. (b) While this error in measurement is not high even at a very high obliqueness angle (70^0), the error is nevertheless present and may account for the error observed in the 3 mm slices.

ing the imaging plane to be perpendicular to the flow direction. Therefore, it is recommended to obtain a suitable scout image before attempting to measure flow velocity in the vessel of interest.

The phase contrast technique provides the best estimate of MR flow velocity, because it contains only information from moving spins[15)] and it is insensitive to most imaging plane has been shown to affect the estimation of flow velocity. The phase contrast technique is sensitive to other artifacts as are most MR techniques to susceptibility, variations in the body surface, inhomogeneity of boundaries of different tissues[14)], respiratory and other kinds of patient motion[10)] and turbulence by irregular inner face and shapes of vessels[27)].

Pulsatile flow may affect oblique flow because pulsatile flow predisposes more to turbulence than constant flow. For the purpose of further investigation on the accuracy of MR flow velocity measurement in the human body, we have plans to study obliqueness of vessel orientation in pulsatile flow phantoms and animals to compare MR measurement results to those obtained with conventional Doppler US examinations, and implanted flow meters in animals. Our study of an in vitro system is a useful step for the clinical application of the flow velocity measurement with the 2D phase contrast technique.

CONCLUSION

The spin phase acquired has been linearly proportional to the flow velocity. Obliqueness of vessels to the imaging plane has been shown to be the most important factor in estimation of flow values. To reduce error, slice thickness should be as thin as possible.

REFERENCES

1) Singer JR : Blood flow rates by nuclear magnetic resonance measurements. Science **130** : 1653〜1653, 1959
2) Schulz RA : MR angiograms detect vascular flow effects. Diagnostic Imaging 111〜116, 1992
3) Axel L : Blood flow effects in magnetic resonance imaging. AJR **143** : 1157〜1166, 1984
4) Axel L, Shimakawa A MacFall J : A time-of-flight method of measuring flow velocity by magnetic resonance imaging. Magn Reson Imaging **4** : 199〜205, 1986
5) Wehrli FW, Shimakawa A, Gullberg GT, MacFall JR : Time-of-flight MR flow imaging: selective saturation recovery with gradient refocusing. Radiology **160** : 781〜785, 1986
6) Shimizu K, Matsuda T, Sakurai, Fujita A, Ohara H, Okamura S, Hashimoto S, Mano H, Kawai C, Kiri M : Visualization of moving fluid: Quantitative analysis of blood flow velocity using MR imaging. Radiology **159** : 195〜199, 1986
7) Evans AJ, Hedlund LW, Herfkins RJ, Utz JA, Fram EK, Vinder RA : Evaluation of steady and pulsatile flow with dynamic MRI using limited flip angles and gradient refocused echos. Magn Reson Imaging **5** : 475〜482, 1987
8) Matsuda T, Shimizu k, Sakurai T, Fujita A, Ohara H Okamura S, Hashimoto S, Tamaki S, Kawai C : Measurement of aortic blood flow with MR imaging: comparative study with Doppler US. Radiology **162** : 857〜861, 1987
9) Evans AJ, Vlinder RA, Herfkens RH, Spritzer CE, Kuethe DO, Frahm EK, Hedlund LW : Effect of turbulence on signal intensity in gradient echo images. Invest Radiol **23** : 512〜518, 1988
10) Edelman RR, Mattle HP, Kleefield J, Silver MS : Quantification of blood flow with dynamic MR imaging and presaturation bolus tracking. Radiology **171** : 551〜556, 1989
11) Moran PR : A flow zeumatographic interface for NMR imaging in humans. Magn Reson Imaging **1** : 197〜203, 1982
12) Van Dijk P : Direct cardiac NMR imaging of heart wall and blood flow velocity. J Comput Assist Tomogr **8** : 429〜436, 1984
13) Moran PR, Moran RA, Karstaed N : Verification and evaluation of internal flow and motion. Radiology **154** : 433〜441, 1985
14) Meier D, Maier S, B siger P : Quantitative flow measurement on phantoms and on blood vessels with MR. Magn Reson Med **8** : 25〜34, 1988
15) Ridgway JP, Smith MA : A technique for velocity imaging using magnetic resonance imaging. Brit J Radiol **154** : 433〜441, 1986
16) Nayler GL, Firmin DN, Longmore DB : Blood flow imaging by cine magnetic resonance. J Computed Assist Tomogr **10** : 715〜722, 1986
17) Firmin DN, Nayler GL, Klipstein RH, Underwood SR, Ree RSO, Longmore DB : In vivo validation of MR velocity imaging. J Comput Assist Tomogr **11** : 715〜756, 1987
18) Underwood SR, Firmin DN, Klipstein RH, Rees RSO, Longmore DB : Magnetic resonance velocity mapping; clinical application of a new technique. Br Heart J **57** : 404〜412, 1987
19) Bogren Hg, Klipstein RH, Firmin DN, Mohiddin RH, Underwood SR, Rees RS, Longmore DB : Quantitation of antegrade and retrograde blood flow in the human aorta by magnetic resonance velocity mapping. Am Heart J **117** : 1214〜1222, 1989
20) Chang JM, Friese K, Caputo GR., Kondo C, Higgins CB : MR measurement of blood flow in the true and false channel in chronic aortic dissection. J Comput Assist Tomogr **15** : 418〜423, 1991
21) Sebok DA, Wilkerson D, Schroder W, Mezrich R, Zatina M : Interleaved magnetic resonance and ultrasound by electronic synchronization. Invest Radiol **26** : 353〜357, 1991
22) Firmin DN, Nayler GL, Kilner PJ, Longmore DB : The application of phase shifts in NMR for flow measurement. Magn Reson Med **14** : 230〜241, 1990
23) Young IR, Hall AS, Bryant DJ, Thimas DGT, Gill SS, Dubowitz LMS, Cowan F, Pennock JM, Bydder GM : Assessment of brain perfusion with MR imaging. J Comput Assist Tomogr **12** : 721〜727, 1988
24) Mohiaddin RH, Amanuma M, Kilner PJ, Pennell DJ, Manzara C, Longmore DB : MR phase-Shift velocity mapping of mitral and pulmonary venous flow J Comput Tomogr **15** : 237〜243, 1991
25) Pettigrew RI, Dannels W : Use of standard gradients with compound oblique angulation for optimal quantitative MR flow imaging in oblique vessels. AJR **148** : 405〜409, 1987
26) Edelman RR, Rubin JB, Buxton RB : Flow. In: Edelman RR, Hesselink JR, eds. Clinical magnetic resonance imaging. Philadelphia: W.B. Saunders Company, 112〜113, 1990
27) Wendt III RE, Nitz WR, Murphy PH : Nuclear magnetic resonance velocity spectra of steady flow. Magn Reson Imaging **15** : 90〜101, 1990

2次元phase contrast法を用いたMRによる流速測定法における撮影面に対する斜め方向の流れがおよぼす測定誤差の評価

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【要旨】 phase contrast法は両極方向に同じ強さの傾斜磁場を作用させることで，遅い静脈血流や脳脊髄液から速い大動脈の流れまで幅広い流速を測定可能であり，また流入および流出両方向の測定も可能である．しかし測定画像断面に対し垂直で無い流れは，その流入角度により誤差を引き起こしていると考えられるが，それを検討した報告はない．そこで測定画像断面に対する液体の流入方向を変化させることにより，引き起こされる測定誤差を評価し検討を行った．

定常流ポンプ，ポリエチレンチューブと流速測定器よりになる実験装置を作成した．1.5TeslaMR装置にて，2次元phase contrast法を用いて実験装置内を流れる生理的食塩水の流速を測定した．流れの角度は測定画像断面に対し垂直から，10度づつ変化させ80度まで測定した．

結果として流入角度は低くなった．またスライス厚が厚い方が誤差は大きかった．従って，2次元phase contrast法による流速測定に関し，できる限りスライス厚は薄くし，測定画像断面に対し流入角度が付いた時には，角度に応じた補正が必要である．

キーワード：Phase Contrast 画像，Oblique flow，磁気共鳴画像，流速測定．