Poststenotic Signal Loss in MR Angiography: Effects of Echo Time, Flow Compensation, and Fractional Echo

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PURPOSE: To evaluate with steady and pulsatile flow the influence of echo time, gradient strength and duration, and flow compensation on the degree of turbulent signal loss, factors that have been implicated in MR angiography's overestimation of the degree of stenosis. METHODS: We examined poststenotic turbulent flow in two models, one that created a turbulent jet and another that simulated a plaque-like stenosis. The pulse sequence used in these experiments allowed for a single variable (flow compensation, echo time, or gradient strength) to be varied without changing the others. RESULTS: Poststenotic signal loss can lead to overestimation of the degree of a stenosis. The area of signal loss in the turbulent jet was influenced by fractional echo and flow compensation, but not by echo time. We found that the dominant mechanism in poststenotic signal loss is related to the strength and duration of the imaging gradients. CONCLUSIONS: Flow-compensated sequences with reduced gradient strength and duration will reduce poststenotic signal loss and may lead to more accurate estimations of the extent of stenotic lesions.

Index terms: Magnetic resonance angiography (MRA); Magnetic resonance, experimental; Arteries, magnetic resonance angiography (MRA); Arteries, stenosis and occlusion

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Magnetic resonance angiography (MRA) has been proposed as a noninvasive alternative to standard contrast angiography as a screening examination for carotid artery vascular disease. Recent work has suggested that MRA compares well both with standard contrast angiography and with duplex ultrasound of the carotid artery (1-5).

One well-recognized problem with MRA is the overestimation of the degree of stenosis (6-9). Turbulent flow past a stenosis leads to signal loss, and this signal loss has been implicated as a major factor in the overestimation of stenosis. For this reason, it would be useful to evaluate the pulse sequence parameters that influence signal loss in the region of a stenosis and to develop pulse sequences that are less sensitive to turbulent signal loss.

This work attempts to evaluate different scanning parameters that may influence signal intensity in poststenotic turbulent flow and aid in designing a pulse sequence that is less sensitive to turbulent signal loss. Using steady and pulsatile flow, we investigate in two different models the effect of echo time (TE), flow compensation (FC), and fractional echo (FE) on the signal intensity of poststenotic turbulent flow.

Materials and Methods

Models

In the first model, we imaged a turbulent jet generated by steady flow through an orifice. The model consisted of a Plexiglas tube, 12 cm in diameter and 40 cm long, with a flat plate in the center of the tube and an orifice 5 mm in diameter in the center of the plate (Fig. 1). Flow emerged in a jet from the orifice. Steady flow through this model was produced with the flow apparatus shown schematically in Figure 1. A reservoir supplied fluid to the model placed in the center of the magnet. The flow rate was regulated by a valve placed in line between the model and the supply
reservoir. Fluid from the model returned to a reservoir and was pumped with a centrifugal pump back to the supply reservoir. An overflow port on the supply reservoir drained fluid from the supply reservoir so that as long as the flow from the pump was greater than the outflow to the model, a constant fluid level and, therefore, a steady pressure head were maintained at all times.

Flow rate through the model was measured by timing the filling of a calibrated beaker and averaging multiple measurements. The pressure drop across the orifice was measured using a simple difference manometer. Flow rates were varied from approximately 100 mL/minute to approximately 4,850 mL/minute. Corresponding pressure differences across the orifice varied from 6 mm to 1810 mm of water.

In the second model, we imaged pulsatile flow past a simulated asymmetric stenosis. A 12.5-mm (inside diameter) Plexiglas tube was fitted with a piece of solid Plexiglas that obstructed 90% of the lumen of the tube (Fig. 2), thus giving a simplified model of a large plaque-like stenosis that might be found in an internal carotid artery.

To generate pulsatile flow, we designed and built a pump, shown schematically in Figure 3. It consisted of a piston that drove fluid forward into a cylinder and out of a one-way valve to the model. On the return stroke of the piston, fluid was drawn through a second one-way valve to fill the cylinder from a storage reservoir, and then was ejected again as the piston moved forward. The piston was connected to a lever that in turn was connected by rod to a flywheel. A ½-hp electric motor (Bodine Electric, Chicago, IL) powered the flywheel via a belt (Fig. 3). This pump generated flow in physiologic ranges, which for these experiments was varied between 125 and 2,100 mL/min.

Blood Simulation Fluid

The viscosity of a fluid affects its behavior during turbulent or disturbed flow. Consequently, a fluid was chosen...
that simulated both the viscosity and the T1 and T2 of human blood. The fluid consisted of 4 parts glycerol to 5 parts distilled water doped with $10^{-4}$ M gadolinium chloride chelated with EDTA. The viscosity of the fluid was 3.8 cPa, the T1 was 800 msec ± 50 msec, and the T2 was 220 msec ± 20 msec. Although hematocrit and viscosity vary from person to person, these values are within physiologic ranges.

**Imager and Scan Parameters**

All imaging was performed on a 1.5 T General Electric Signa System (Milwaukee, WI). Three different parameters were independently varied and evaluated: TE, FC, and FE. Parameters were: repetition time (TR), 27 msec; TE, 3.2 to 20 msec; matrix, $128 \times 256$; two excitations; $30^\circ$ flip; field of view, 20 cm; and a two-dimensional Fourier transform technique.

The TE is defined as the time between the radiofrequency pulse and the echo center when the received signal is largest. Typically, the echo center coincides with the center of the readout gradient, and the received signal is symmetric. A FE (asymmetric) may be used if the homodyne reconstruction technique is applied to the individual echoes. This is similar in concept to fractional number of excitation imaging, but the asymmetry occurs in the individual echoes rather than in the acquired phase encodings (10). FE was defined such that a perfectly centered echo was given a value of 100% while 50% was assigned to "half echo" or free induction decay (FID) starting at the beginning of the readout window (Fig. 4). While the readout gradient amplitude is not affected, its duration prior to the echo (the gradient lobe that determines the position of the echo in the frequency-encoding gradient) center is reduced with the symmetry fraction. Because the prephaser and the FC gradient lobes are calculated according to the strength of the readout gradient and its duration prior to the echo center, their strength and duration are reduced for asymmetric echoes. Dephasing due to uncompensated motion (acceleration, jerk, etc., for flow-compensated sequences, and for all motion when the FC is not used) is minimized by playing out the prephaser and FC gradients immediately prior to the readout regardless of TE or echo asymmetry.

**Image Analysis**

Images were analyzed with the purpose of evaluating the effect of changing one of the scan parameters (TE, FE, FC) on the appearance of the poststenotic turbulent flow. With the model that produced a jet through an orifice, the jet resulted in a region of decreased signal intensity.

The area of the jet in square millimeters was measured in a consistent way. The image was windowed with the width set at 1 and the level at a value that represented one-half the maximum signal intensity in the image. An example of this technique is shown in Figure 5. This method produced a reproducible and easily identifiable edge and accounted for differences in the signal intensity in the fluid. A cursor was traced around the edge of jet and the area was calculated. This technique of area measurement was reproducible ±6%.

**Clinical Studies**

After informed consent, using a protocol approved by an independent review board, two patients with moderate to severe stenoses of the internal carotid arteries underwent two-dimensional time-of-flight MRA; TR was 13 msec, TE 4 msec, flip angle was 45°, slice thickness was 3 mm, matrix $128 \times 256$, and field of view 18. Standard clinical imaging parameters were used, with the exception that one
set of images was acquired with a FE and one set was acquired with a full echo. Postprocessing and three-dimensional reprojections were obtained using a maximum intensity pixel algorithm.

Results

Keeping the TR constant at 27 msec, TE was varied from a minimum of 3.2 msec to a maximum of 20 msec. Figure 6 demonstrates the effect of changing TE on the area of signal loss in the turbulent jet. In all these experiments, the flow was constant at a rate of 1950 mL/minute and the difference in pressure across the orifice was 30.2 cm of water. Only the TE was changed; the other scan parameters (FE, FC, TR) were held constant. The experiment was repeated with and without FC and at FEs of 51% and 100% (full echo), and similar results were obtained. The area of signal loss varies little as TE changes, as seen in Figure 6. Examples of the actual images corresponding to these data are shown in Figure 7. As shown in Figures 8 through 10, FC does have an effect on the area of signal loss. The images shown were acquired at identical flow rates and at the same TE, but with and without FC and at a fractional or a full echo. Other parameters being equal, images acquired with FC demonstrate less signal loss than those without (Figs. 8–10). Signal loss is least with a FE and FC, and greatest with a full echo and no FC.

Although TE has little effect on the area of signal loss, the echo fraction does have an effect. This effect is demonstrated in Figures 10 and 11. TE and flow rate are constant; only the echo fraction (percent full echo) changes. In these experiments, the echo fraction varied from 51% to 100% (full echo). A FE of 51% produces the least signal loss, and the area of signal loss increases as the echo becomes less fractional, so that the most signal loss occurs with a full echo.

With increasing flow rate (and therefore, increasingly disturbed flow) through the orifice, the pressure difference across the orifice increased. As the pressure difference increased, the area of signal loss also increased (Fig. 12). The point at which signal loss first occurred depended on the fractionality of the echo and the presence or absence of FC. For instance, at a FE of 51%, signal loss first occurred at a pressure difference of approximately 90 mm of water, whereas at a full echo, signal loss first occurred at a pressure difference of 10 mm of water. The presence of FC affected signal loss more at a full echo than...
Fig. 9. Images of poststenotic signal loss at the same flow rate (1950 mL/minute) with varying FE and FC. On the upper row of images from left to right, the first image represents a FE of 51% with FC, the second image a FE of 51% without FC, the third image a full echo with FC, and the fourth image a full echo without FC. The images below are the same except windowed at the width and level used to calculate area of signal loss.

Fig. 10. Area of signal loss versus FE. Data were acquired with a constant flow velocity of 1950 mL/minute. ■, data acquired with FC; □, data acquired without FC. TE is constant at 10.2 msec.

Fig. 11. Areas of poststenotic signal loss obtained at a constant flow rate, constant TE, and with FC, but varying FE. The image on the upper left is a FE of 51%, upper middle FE = 60%, upper right FE = 70%, lower left FE = 80%, lower middle FE = 90%, and lower right full echo.

with a FE. Interestingly, with a full echo and no FC, signal loss first occurred at a pressure difference of 10 mm of water, but with FC and a full echo, it did not occur until a pressure difference of approximately 30 mm of water was reached.

One element of the pulse sequence that changed with FE was the strength and the duration of the gradients used to generate the echo. In order to have a simple parameter that reflected the strength and the duration of all the imaging gradients, we chose to calculate the absolute value of the area of all the gradients between the time of the RF application and the echo center. We call this parameter the absolute gradient area. Figures 13A and 13B show the area of signal loss as related to the absolute gradient area for each image. Signal loss increased as gradient strength and duration increased. In the non-FC case, it is clear that as the echo fraction increased, the first and higher moments also increased while in the FC case, only moments of higher order than linear increased. Note that with FC, the absolute gradient areas were higher than without FC, but the trend was the same; the signal loss still increased with absolute gradient area.

The absolute values of the gradients were calculated also for change in TE. The gradient strengths do not change with change in TE.

Images from the stenosis model are shown in Figures 14A and 14B. Pulsatile flow rate was 250 mL/minute at 60 beats per minute. Signal loss reached its maximum at the FE of 100% (full echo). Note that the signal loss occurred both poststenosis (in the "jet" downstream) and actually at the stenosis. Figure 14B shows a magnified view of the stenosis itself. At a FE of 51%, flow through and past the stenosis was well visualized. At a FE of 100%, no signal was seen from flow through the stenosis. Note also that
signal loss occurred actually before the stenosis in the full echo image, but not with a FE of 51%.

Images from two patients with moderate to severe stenoses are shown in Figure 15. The MRA are acquired with FC, and at a FE of 51% and at a full echo. In both cases, the FE of 51% demonstrated less signal loss poststenosis and gave a more accurate representation of the actual appearance of the stenosis compared with the conventional contrast angiograms. In both cases, there was a complete loss of signal in the region of stenosis with the images acquired at a full echo, while no such signal loss occurred with a FE of 51%. The images acquired with the full echo could be falsely interpreted as a complete occlusion, while those acquired with the FE demonstrate that the lumen is patent.

**Discussion**

The effects of TE, FC, and FE on the signal loss in a turbulent jet have been investigated in this paper. That turbulent flow can produce signal loss in poststenotic flow has been previously demonstrated (11, 12). That this signal loss can lead to overestimation of stenosis has been asserted, but, to our knowledge, has not been systematically proven (6, 7). Our work supports the idea that poststenotic signal loss may lead to overestimation of stenotic lesions (Figs. 14 and 15). Further, the evidence suggests that FC and FEs affect the appearance of poststenotic signal loss (Figs. 14 and 15).

It is a commonly expressed notion that decreasing TE should decrease poststenotic signal
loss (8, 9). The results of this paper suggest that this is an oversimplification. Over the range of TEs examined, TE did not significantly influence the area of signal loss in a turbulent jet. In these experiments, shortening TE only decreased poststenotic signal loss when a FE was used to shorten the TE. That TE could affect signal loss at shorter TEs is a possibility we could not investigate.

The acquisition of a FE does influence poststenotic signal loss. As the fraction of the acquired echo decreases, signal loss poststenosis also decreases. This phenomenon occurs with and without FC, and independent of TE (Figs. 8 and 10). A parameter that changes with FE is the absolute gradient area. This value reflects the strength and duration of the gradients used to produce the image, and changes signal loss (Fig. 13), thus supporting the idea that an increase in gradient strength and duration affects poststen-
otic signal loss. Urchuk and Plewes have recently reported a similar dependence of signal loss on the duration of imaging gradients (13).

First-order FC also affects signal loss poststenosis. At the threshold at which signal loss occurs, images with FC show no signal loss, whereas images without FC show significant signal loss. Likewise, a smaller area of signal loss occurs in images with FC than in those images taken with the same parameters without FC (Fig. 10). The effect of adding FC is more prominent at a full echo (greater gradient strength and duration) than at a FE (lesser gradient strength and duration), suggesting that FC may have less effect on poststenotic signal loss in a pulse sequence that uses very low gradient strengths to create the image.

Previous investigations of flow-related artifacts have centered either on the effect of TE or on the effect of gradient strength and duration. Some have emphasized the reduction of TE as a means of reducing these artifacts (including poststenotic signal loss) (8, 9, 14). Others have recognized that decreasing gradient strength and duration can reduce motion artifacts and signal loss in regions of disturbed flow (10, 13, 15–17).

In this study, we have examined these effects separately, and have investigated the effect of FC. Our results suggest that there is an element of poststenotic flow that is influenced by FC, and there is an element that is affected primarily by gradient strength and duration and not by TE. None of our results support the previously expressed idea that TE influences poststenotic signal loss. If turbulence is a dephasing phenomenon, then perhaps the dephasing occurs on such a short time frame that TEs on the order of those used in conventional imaging do not affect it.

These experiments did not address the effect of decreasing voxel size on turbulent signal loss. It is possible that three-dimensional sequences that use smaller voxel sizes may partially offset intravoxel dephasing; we did not investigate this phenomenon.

Practically speaking, it seems reasonable to expect that pulse sequences designed with FEs and FC would be less sensitive to poststenotic signal loss, and, therefore, may lead to a more accurate estimation of stenosis. Likewise, TE can be adjusted to suit the implementation of these two parameters and to optimize signal-to-noise and contrast-to-noise. A logical extension of these ideas is to use FID imaging, which minimizes gradient strength and duration. In these experiments, the effect of FC was lesser when the gradient strengths were reduced (FE imaging). This suggests that FC may be unnecessary (as well as impractical) with FID imaging.

Conclusions

Three main conclusions may be drawn from this work. First, poststenotic signal loss can lead to overestimation of the degree of a stenosis. Second, the dominant mechanism in poststenotic signal loss is related to gradient strength and duration rather than TE. Third, FC reduces poststenotic signal loss, but this effect is less prominent in images acquired with lower gradient strength and duration. Finally, it seems reasonable to expect that flow-compensated sequences with reduced gradient strength and duration will reduce poststenotic signal loss and may lead to more accurate estimations of the extent of stenotic lesions.

References

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