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Spatial Misregistration of Vascular Flow During MR Imaging of the CNS: Cause and Clinical Significance

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Spatial misregistration of signal recovered from flowing spins within vascular structures is a common phenomenon seen in MR imaging of the CNS. The condition is displayed as a bright line or dot offset from the true anatomic location of the lumen of the imaged vessel. Its origin is the time delay between application of the phase- and frequency-encoding gradients used to locate spins within the plane of section. The principal condition necessary for the production of spatial misregistration is flow oblique to the axis of the phase-encoding gradient. Flow-related enhancement (entry slice phenomenon), even-echo rephasing, and gradient-moment nulling contribute to the production of the bright signal of spatial misregistration.

Familiarity with the typical appearance of flow-dependent spatial misregistration permits confirmation of a vessel's patency; identification of the direction of flow; estimation of the velocity of flow; and differentiation of this flow artifact from atheromas, dissection, intraluminal clot, and artifacts such as chemical shift.

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Vascular flow effects give rise to artifacts and variable intravascular signal intensity during MR imaging. As a noninvasive tool, MR imaging delineates the morphologic characteristics of the vascular system and provides insight into the physiological parameters of blood flow.

Several authors have reported spatial misregistration of flowing blood during routine spin-echo (SE) MR imaging [1–6]. The typical appearance of spatial misregistration as a bright line or oval adjacent to an imaged vessel is a common finding during MR imaging of the CNS. This article presents the theoretical basis for the misregistration artifact and how this knowledge can be used to determine vessel patency and direction of blood flow, to estimate its velocity, and to differentiate misregistration artifact from other artifacts and pathologic vascular processes.

Materials and Methods

MR scans of the head were reviewed to identify representative instances of spatial misregistration of intracranial vascular flow.

MR examinations were carried out on a 1.5-T MR unit (General Electric, Milwaukee, WI). Slice thickness was 3 or 5 mm and an intersection gap of 20% or 50% was used. The phase-encoding matrix was 128 or 256.

Most patients underwent short TR/short TE, 400–600/20–25/2 or 4 (TR/TE/excitations), and long TR/short and long TE, 2000/25–80/2, evaluations without flow-compensation gradients. Five patients were studied with T2-weighted examinations (2800/30,70/2) in which flow-compensation gradients were applied along the readout and slice-selection gradient axes [7–9]. One patient was examined with three axial T2-weighted scans consisting of a routine 2000/35,70/2 study, a four-echo 2000/25,50,75,100/2 sequence to generate even-echo rephasing, and a 2000/35,75/2 protocol in which the directions of the phase- and frequency-encoding gradients were exchanged relative to the routine study. One normal

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volunteer was examined first with 2000/40/2 and a second time with 2000/80/2, both studies using an SE sequence with flow-compensation gradients. Five patients were studied with a multiplanar gradient-recalled echo (MPGR) technique using 250–600/14–19/4 and flip angles of 35–60°. One patient was studied with sequential slice gradient-recalled echo acquisition in the steady state (GRASS) [7, 9, 10] using 100/12 and flip angle of 10°. Flow-compensation gradients were used with both the MPGR and GRASS sequences. Cardiac gating techniques were not used. The results of this study were considered to be representative of flow imaged randomly during the cardiac cycle.

The predicted appearance of spatial misregistration was tested by exchanging axes of the phase- and frequency-encoding gradients, by varying the TE after phase encoding, and by using a symmetric four-echo sequence. Because spatial misregistration was commonly demonstrated in venous angiomas, seven examples in six patients were included and analyzed. Several venous angiomas were examined further with angiography for confirmation of the vascular anomaly and its drainage pattern.

The velocity of vascular flow was estimated in seven venous angiomas by using SE series with even-echo techniques, in five convexity cortical veins in five different patients by using SE MR with flow-compensation gradients and the shortest TE value (30 msec) in a multiecho sequence for velocity calculations, and in five intracranial arteries (one vertebral, one proximal anterior cerebral, one proximal posterior cerebral, one proximal middle cerebral, and one distal middle cerebral) in five patients by using MPGR imaging. Because velocity (v) equals distance traveled (d) divided by time, knowledge of the time interval between application of the phase and readout gradients permits estimation of flow velocity:

$$v = d/(TE - t_{phase}): \quad (1)$$

where t_{phase} is the time of the phase-encoding gradient application. Equation 1 was applied to examples of in-plane flow oblique to the axes of the phase- and frequency-encoding gradients. Calculations were based on measurement of the hypotenuse of a right triangle constructed with sides parallel to the phase- and frequency-encoding gradient directions and the hypotenuse lying along a straight segment of the imaged vessel (Fig. 1). This calculation provides the approximate velocity a flowing spin traveled between phase-encoding and readout steps.

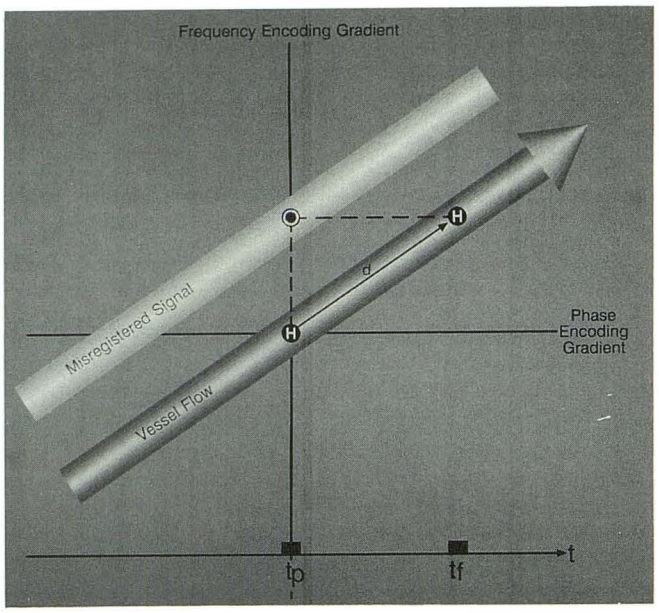

Fig. 1.—Diagram of in-plane vessel obliquely oriented to phase- and frequency-encoding gradients and geometric basis for derivation of equation 1 (see text). H = flowing spin, O = point of misregistered signal, d = measured distance traveled by flowing spin, t = time axis, t_p = time of phase-encoding gradient application, t_f = time of frequency-encoding gradient application.

Results

Spatial misregistration of vascular flow was routinely identified in all three standard MR imaging planes (sagittal, coronal, axial) used to examine the head. With the use of even-echo rephasing during a symmetric, multiecho SE sequence, bilateral cerebellar angiomas exhibited spatial misregistration on the second and fourth echoes of a four-echo series (Fig. 2). The misregistered line of the fourth echo was spatially mismapped a greater distance from the angioma than on the second echo.

The spatially misregistered bright line was found to be displaced on opposite sides of two different venous angiomas depending on the direction of their blood flow (Fig. 3). In each case, the spatially misregistered signal was displaced along that portion of its flow, which was directed along the frequency-encoding axis and mismapped upstream along the phase-encoding axis. Exchanging the orientations of the phase- and frequency-encoding gradients reversed the location of the bright line relative to the true lumen of a cerebellar venous angioma (Fig. 4).

The center of the spatially misregistered signal was always displaced the same distance from the imaged vascular structure when flow-compensation gradients were used regardless of TEs chosen. This was demonstrated on multiecho SE sequences and in the volunteer who underwent a head examination with two different TEs. Cortical veins imaged in the presence of flow-compensation gradients (Fig. 5) illustrated that the width of the spatially misregistered signal is of equal or slightly increased width with longer echo delays, the measured distances traveled by a flowing spin on both a single short TE/long TR and single long TE/long TR studies were approximately equal, and velocity calculations became dependent on the TEs selected.

Equation 1 was applied to estimate velocity of in-plane flow in venous angiomas and in normal intracranial veins and arteries. Flow rates were calculated to be 2.3–5.8 cm/sec (±0.21 cm/sec) in venous angiomas, 11.7–26.4 cm/sec (±0.35 cm/sec) in large cortical veins, and 26.8–42.6 cm/sec (±0.75 cm/sec) in intracranial arteries.

Discussion

The appearance of a bright line or oval adjacent to a CNS vascular structure is common during MR imaging. When an imaged vessel lies within the plane of section oblique to both the phase- and frequency-encoding gradients, spatial misregistration is represented as a bright white line of approximately uniform width offset slightly from the true anatomic

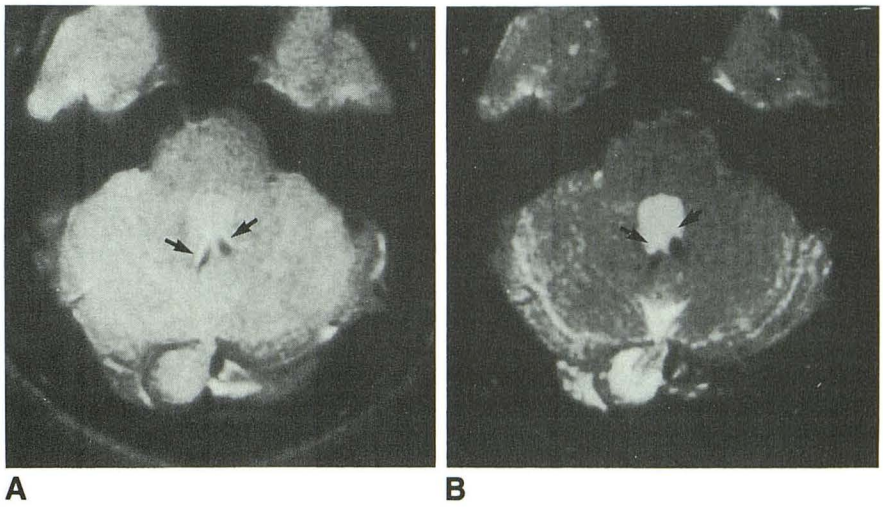

Fig. 2.—*A* and *B*, Axial second, SE 2000/50 (*A*), and fourth, SE 2000/100 (*B*), echoes of T2-weighted four-echo series (SE 2000/25,50,75,100) of the head. Centrally draining bilateral cerebellar venous angiomas show spatial misregistration (*arrows*) on second echo, which is further displaced from parent vessel on fourth echo, becoming indistinguishable from white CSF of fourth ventricle.

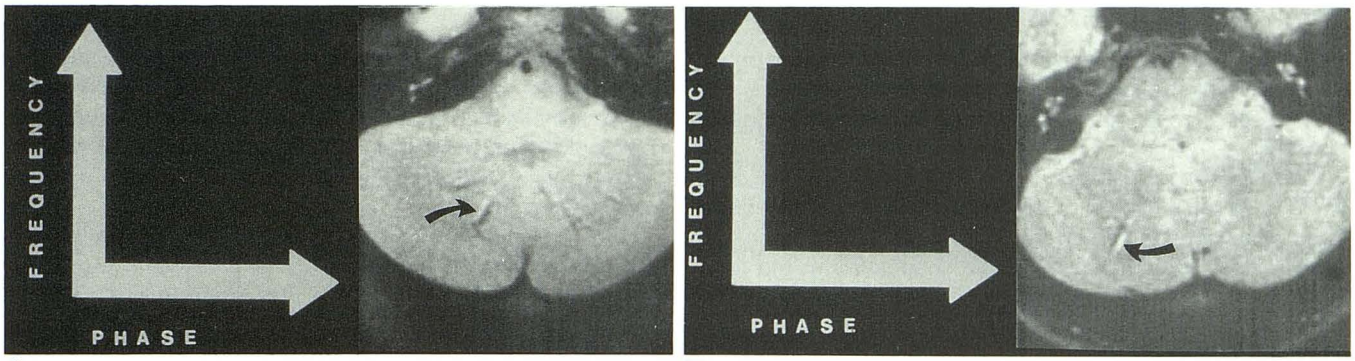

Fig. 3.—Axial T2-weighted second-echo images of cerebellum in two patients with venous angiomas.
A, Venous angioma and its accompanying bright misregistered signal (*arrow*) predict venous anomaly drains centrally.
B, Cerebellar venous angioma in this patient drains peripherally, and misregistered signal (*arrow*) is on opposite side of anomalous draining vein when compared with example in *A*.

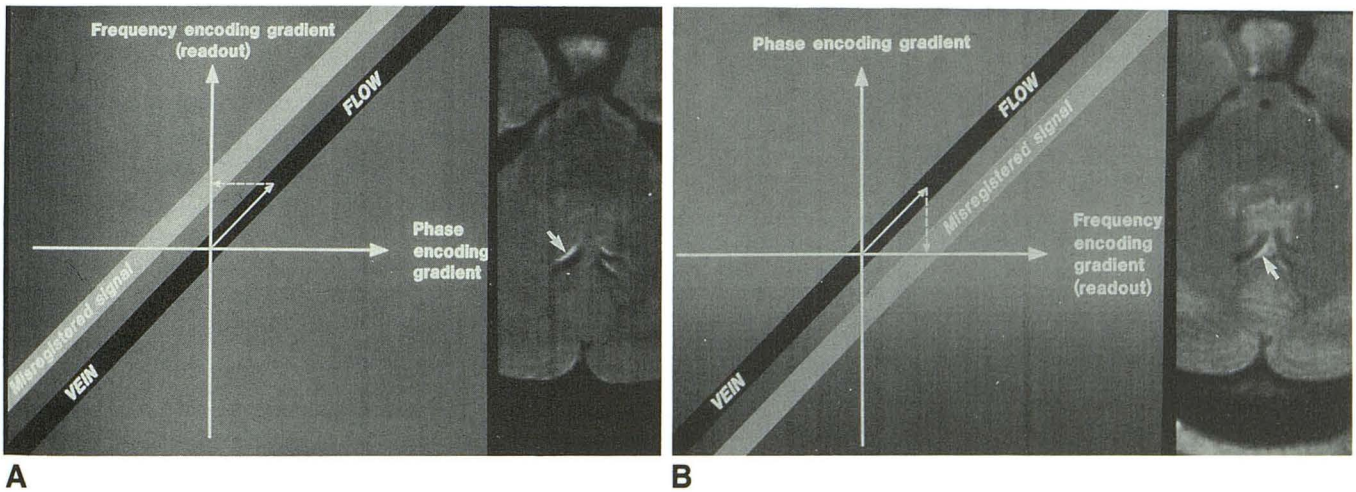

Fig. 4.—*A* and *B*, A patient with bilateral cerebellar venous angiomas was examined first with usual application of phase- and frequency-encoding gradients and then with axes of spatial-encoding gradients swapped. Second echo of each T2-weighted axial study shows spatially misregistered lines (*arrows*) to change their relationship to parent vessels depending on orientation of phase- and frequency-encoding gradients. Accompanying diagrams illustrate that misregistered signal of right cerebellar venous angioma is displaced along direction of its flow, which lies along frequency-encoding direction (upstream).

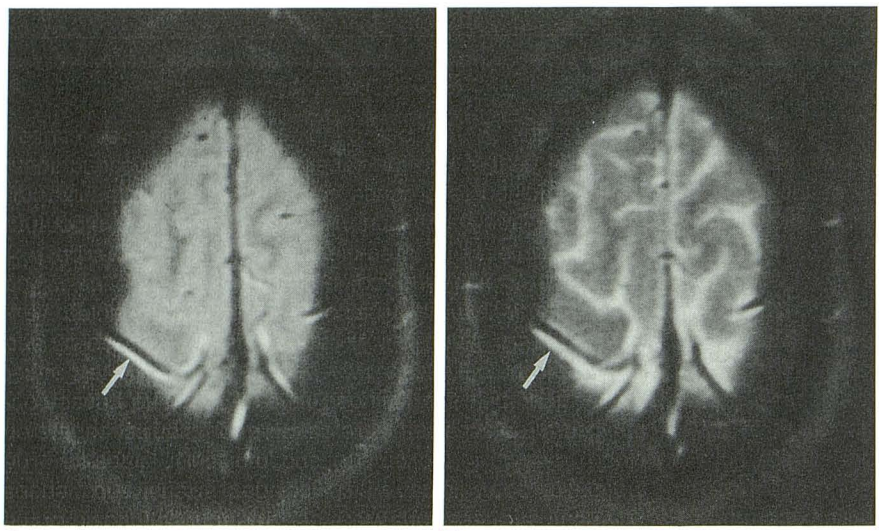

Fig. 5.—*A* and *B*, First- (*A*) and second- (*B*) echo T2-weighted axial images of head using flow-compensation gradients. Misregistered signal (*arrows*) occurs adjacent to normal superficial cortical draining veins on all echoes and is of equal width and misregistered equal distance from true vessel. Misregistered signal is more apparent on first-echo image.

location of the imaged vessel. This can be termed upstream spatial misregistration because the signal intensity of the moving spin is mapped at a more proximal position along the phase-encoding axis.

Downstream mismapped signal of flowing spins occurs when a vessel is parallel to the frequency-encoding gradient axis or the vessel traverses the plane of section. Spin position under these circumstances is determined solely at the time of frequency encoding as long as phase dispersal of the flowing spins has not occurred and there is no flow in the phase-encoding direction.

When the vessel is in plane and either aligned with the phase- or frequency-encoding gradients, the spatial misregistration artifact is difficult or impossible to identify because the mismapped signal is aligned with the vascular structure and has no lateral offset from the lumen of the parent vessel. Flow within a vessel coursing along the direction of the phase-encoding gradient causes signal from moving spins to be positioned upstream in line with the vessel's lumen. Flow within a vessel perpendicular to the phase-encoding gradient causes signal from moving spins to be located downstream, but again in alignment with the true vessel lumen. For spatial misregistration of vascular flow to be appreciated, the vessel examined must be oriented oblique to the axis of the phase-encoding gradient.

When the vessel traverses obliquely through the plane of section, the resultant misregistered signal is manifested as a dot or oval of high intensity adjacent to the imaged vessel and is always mismapped downstream. This resembles partial-volume averaging of the parent vessel and its misregistered signal when the three-dimensional section is transformed into a two-dimensional image (Fig. 6). A vessel perpendicular to the plane of section will produce no spatial misregistration. When signal intensity from the imaged intraluminal spins is present, it will be correctly placed within the vessel of origin when the vessel courses at a 90° angle through the plane of section (Fig. 7).

The bright white line or dot that appears adjacent to a vessel is explained as the signal of flowing blood spins assigned to pixel locations outside of the true vessel and superimposed on adjacent anatomy. The signal intensity from these spins is then summed with the signal of the stationary spins of the adjacent tissue. The vessel of interest demonstrates signal void because few spins are spatially mapped within its lumen.

The fundamental cause of the misregistered signal is the interval between the applications of the phase- and frequency-encoding gradients used during two-dimensional Fourier localization of spin signal intensity [1, 5, 6]. For the MR system used in this study, a phase-encoding gradient is applied 1.6 msec after delivery of the 90° nutation RF pulse to encode spins along one axis. This time interval is not altered by varying the TE. The frequency or readout gradient is applied later at time TE after the spins have traveled downstream (Fig. 8). The flowing spin first is given a phase coordinate identity and then downstream is read out with a frequency labeling process. The two coordinates for phase and frequency describe a point in a two-dimensional matrix and locate a position for the spin in the two-dimensional Fourier-transform reconstructed image.

Through-plane flow compounds this explanation of spatial misregistration by additionally contributing the time delay from the preceding RF pulse experienced by a slice-selected flowing spin to subsequent readout as another determinant of spin localization. This is the mechanism by which through-plane flow is always mismapped downstream.

The greater the time interval between the applications of the phase- and frequency-encoding gradients, the greater the distance the spatial misregistration is displaced from the parent vessel [5, 6]. This is evident in SE studies (without flow-compensation gradients) when comparing late-echo with early-echo spatial misregistration because the echoes are read out at progressively longer intervals from the application of the single phase-encoding gradient early in the timing sequence.

The longer the TE (time to readout), the more time is available for spins to dephase [5, 11]. Flowing spins with excessive dephasing or turbulent flow will not produce spatial

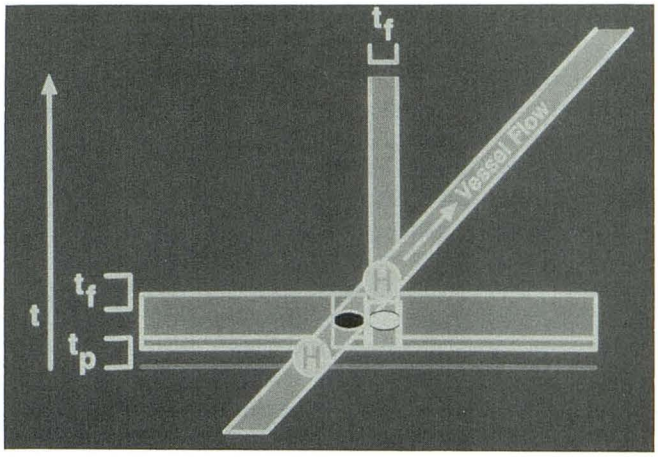

Fig. 6.—Schematic illustration of vessel that obliquely traverses plane of section and its downstream misregistered signal. Acquired slice is viewed in cross section with finite width. Flowing spin (H) experiences first phase-encoding gradient, t_p, then frequency-encoding gradient, t_f. Spatial-encoding gradients are oriented within plane of slice or perpendicular to its cross section. Frequency-encoding gradient application, t_f, is shown twice to illustrate both time elapsed between two localization gradients and their orthogonal relationship. t = time.

misregistration because signal intensity is reduced owing to the destructive interference of spins with many different phase angles [3, 5, 6]. Fundamental causes of flowing spin dephasing are the phase-encoding, slice-selection, and frequency-encoding gradients, which create phase shifts [1, 3, 6, 9, 11–21] dependent on the position of each spin within the intraluminal flow profile [1, 6, 12, 14]. This leads to intravoxel phase dispersion [1, 5, 6, 11, 12, 14, 17, 18] and phase modulation resulting in diminution of pixel signal amplitude [4, 12, 15, 19] for a wide distribution of phase angles [6, 13, 17], as well as to the commonly encountered phase-encoding artifact due to bulk motion, which generates a narrower range of phase shifts [6, 13, 17]. Even-echo rephasing [3, 5, 11] or flow-compensation gradients [6–9, 17] compensate for velocity-induced phase shifts.

Even-echo rephasing in slow flow (predominantly constant velocity) causes the majority of spins in the presence of balanced gradients to be in phase on even-numbered echoes [1, 3, 6, 11, 14–16, 19, 22]. Rephasing of spins is not always complete on even echoes when more complex flow is present involving acceleration or jerk [5, 11] and can occur on odd echoes, particularly with pulsatile flow [20].

Flow-compensation gradients (gradient moment nulling) attempt to correct for spin dephasing that is generated by gradients applied to differential velocity profiles [1, 6, 12, 14] normally found in intracranial vessels. Biphasic flow-compensation gradients are enabled during slice-selection and readout gradient applications. Rephased flowing spins become spatially misregistered with each TE because of the use of flow-compensation gradients with each readout. With flow-compensation gradients, each echo results in misregistered signal of equal width displaced an identical distance from the parent vessel. Flow-compensation gradients cannot compensate for phase shifts associated with flow of nonconstant velocity (due to acceleration and jerk) [7] nor for frequency shifts of flowing protons that are generated in the presence of gradients [21]. This latter effect may cause some of the blurred margins of the spatially misregistered signal and may be one of the mechanisms responsible for the roughly identical appearance of the misregistered signal on any echo when flow-compensation gradients are employed. Under the influence of flow-compensation gradients, spatial misregistration is more apparent on the first than on the second echo of an SE sequence (Fig. 5). By the time of the second echo, spin dephasing and washout effects diminish signal intensity, and the detection of the misregistered signal is more difficult against a background of bright CSF (Figs. 2 and 5B).

Slow flow, as is commonly present in intracranial veins and dural sinuses, produces spatial misregistration during SE imaging without flow-compensation gradients. If the vascular flow is very slow, no spatial misregistration will occur, or the mismapped signal will overlap the walls of the vessel imaged. Typical parabolic velocity flow profiles (e.g., venous) due to laminar flow [4, 5, 23] would cause the fastest flowing spins to be displaced greatest from the parent vessel lumen, slower flow the least, resulting in a band of mismapped parallel signal. Arterial flow, however, is nonlaminar, disturbed, pulsatile, and pluglike [1, 5]. Pulsatile flow has been shown to reverse the intraluminal velocity gradient with slower flow centrally and faster flow peripherally [23]. This should contribute greater width to the spatially misregistered signal band.

During spin-echo sequences, flowing spins must experience both the 90° and the 180° RF pulses to contribute signal intensity [3, 5, 11, 18, 22]. When the imaged vessel

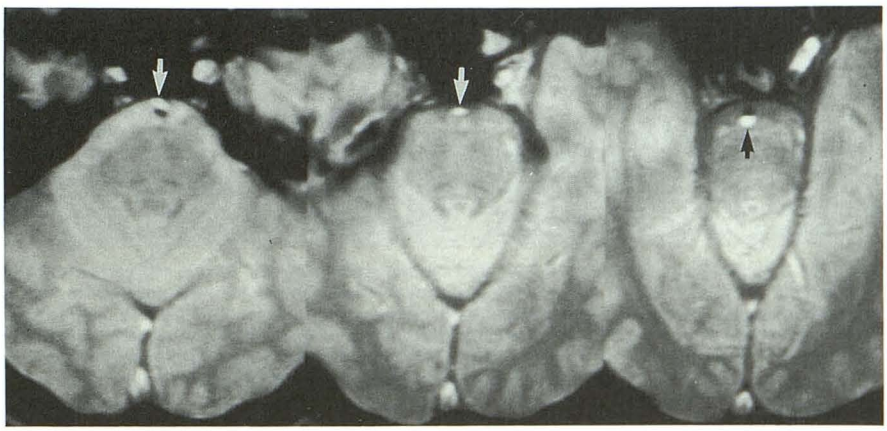

Fig. 7.—Three consecutive axial images of head using multiplanar gradient-recalled echo techniques. Basilar artery traverses axial plane and, depending on obliquity of its course through plane of section, oval or round dots of spatial misregistration (*arrows*) are mismapped anteriorly (left image) or posteriorly (right image), or not misregistered (central image).

courses orthogonal to the plane of section, the spin's flow velocity determines whether the flowing spins will experience the 90° and the 180° slice-selective pulses and the phase-encoding and readout gradients within the appropriate slice. Flow must be slow enough in a typical SE study for the bright dot or oval to be visibly registered next to the round signal void of the imaged vessel. At rapid flow rates, high-velocity washout effects [5, 11] and phase dispersal due to spin-velocity gradients [5, 9, 12] diminish or eliminate the signal source for spatial misregistration during routine SE imaging [3], particularly with longer TEs [5, 11]. However, fast flow, present in arteries and larger venous intracranial vessels, will produce spatial misregistration when gradient-recalled echo techniques are used (Fig. 9) [11].

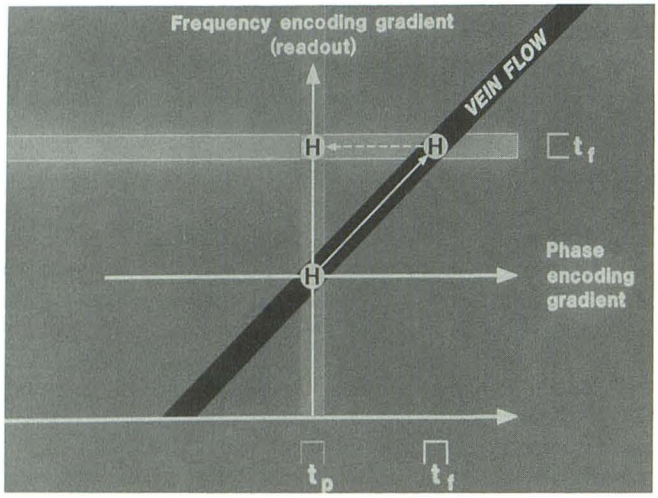

Fig. 8.—Simplified diagram of solitary spin traveling within vessel that is oblique to phase- and frequency-encoding gradients and in plane. Spin is first phase encoded and then frequency encoded, causing it to be localized outside parent vessel, upstream with respect to phase-encoding gradient axis. H = flowing spin, t_f = time of frequency-encoding gradient application, t_p = time of phase-encoding gradient application.

With gradient-recalled-echo imaging, the flowing spins must experience the initial nutation pulse, the phase-encoding step, and the refocusing readout gradient to produce MR signal. In this sequence a slice-selective 180° pulse is not used. Therefore, the signal of flowing spins can be spatially frequency encoded by using a gradient reversal technique (gradient applications are inherently nonselective), even though they have left the slice section at the time of readout [13]. This allows for signal generation from more rapid flow [5, 9, 13, 18, 24]. Flow-related enhancement is another significant source of MR signal from flowing spins during gradient-recalled-echo imaging [9, 13], particularly when slices are acquired sequentially.

Flow-related enhancement (entry slice) or even-echo rephasing are requisites for the production of spatial misregistration using routine SE MR techniques without flow-compensation gradients. Flow-related enhancement phenomena have been explained in detail by other authors [1, 5, 11, 14, 25]. They are also relevant when an interslice gap is interposed between slices, allowing more fully magnetized spins to enter the partially saturated plane of section and emit a strong echo. Flow-related enhancement in normal venous structures often results in spatial misregistration when short TE/short TR SE images are produced, especially in or near the end slices of a multislice sequence (Fig. 10).

By examining the relationship of the spatially misregistered signal to the linear signal void of the corresponding vessel and knowing the orientation of the phase- and frequency-

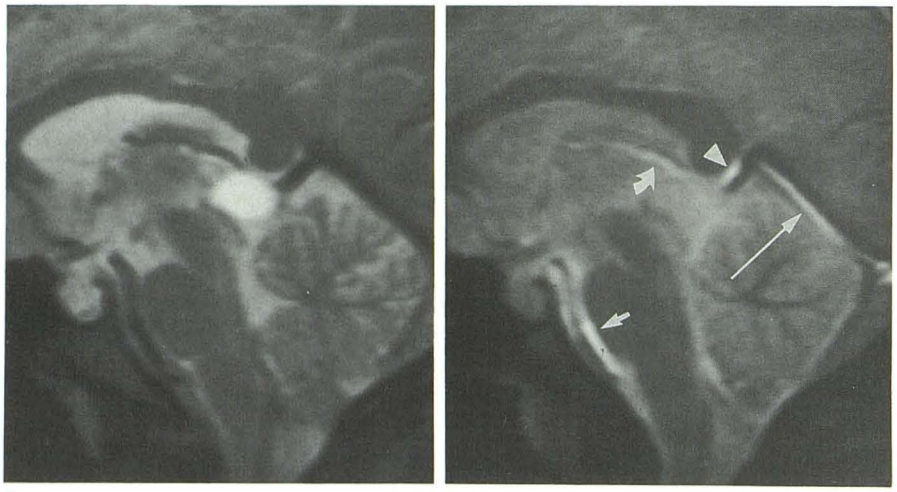

Fig. 9.—*A* and *B*, Second echo of T2-weighted sagittal study, SE 2000/70 (*A*), and sagittal GRASS examination, 100/12/10° (*B*), in patient with pineal cyst. Spatial misregistration adjacent to basilar artery (*short straight arrow*), internal cerebral vein (*curved arrow*), great vein of Galen (*arrowhead*), and straight sinus (*long straight arrow*) is apparent on GRASS scan. Note that misregistered signal is correctly placed depending on direction of flow within vessel and changes its orientation to parent vessel depending on vessel's obliquity to phase- and frequency-encoding gradients (compare great vein of Galen with straight sinus).

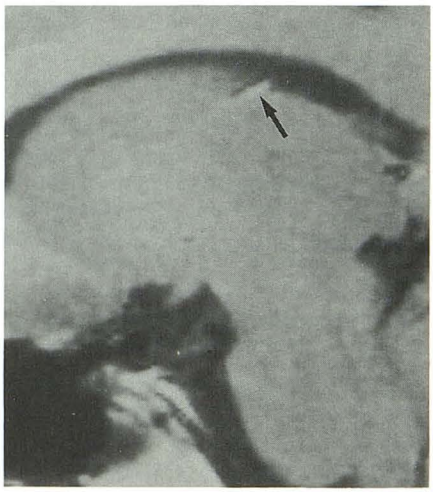

Fig. 10.—Parasagittal short TR/short TE SE image. Flow-related enhancement in left thalamostriate vein causes spatial misregistration, seen as bright white line (*arrow*) adjacent to black line of imaged vessel.

encoding gradients with respect to the plane of section, the direction of flowing blood can be determined (Fig. 11) [5]. In the most common situation, where the vessel is obliquely oriented to the phase- and frequency-encoding gradients, its directionality can be broken down into two vectors, one vector along the axis of the frequency-encoding gradient and the other along the axis of the phase-encoding gradient. This creates a right triangle with its hypotenuse along the imaged vessel. The corner of the right angle of this triangle marks the point of misregistered signal derived from a flowing spin (Fig. 1). The use of spatial misregistration to predict the direction of flowing blood is a practical aid to facilitate the correct distinction of normal patent arteries and veins or when evaluating supply and drainage of pathology with abnormal vascularity.

The relationship of the bright line or dot to the imaged vessel depends on the orientation of the phase- and frequency-encoding gradients for each particular plane of section and can be controlled through settings of operator preference. Different manufacturers orient the phase- and frequency-encoding gradients in various slice planes differently.

The velocity of in-plane flowing blood can be estimated through the use of simple geometry. The results of flow calculation in this study indicate that an approximation of venous or arterial flow can be achieved. The calculations for intracranial arteries are in general agreement with measurements of flow velocities in the middle cerebral, anterior cerebral, and posterior cerebral arteries by transcranial Doppler methods [26, 27]. Measurements of venous flow rates in cortical veins or venous angiomas are not available in the literature. Difficulties with the use of spatial misregistration to determine flow velocities lie in the tedious collection of the velocity data with calipers and in the potential error introduced when recording millimeter measurements. Additionally, the use of flow-compensation gradients causes equation 1 to become dependent on the TE values chosen for velocity calculation since the measured distance of a moving spin is approximately equal for each echo.

While any tubular flow in the CNS can produce spatial misregistration, such as that associated with flowing blood or CSF, recognition of this phenomenon is important in order to distinguish it from other artifacts or true disease. There should be no confusion with atheromas, dissections, or clotted blood, which have different appearances and cannot be altered by changing the directions of the phase- and frequency-encoding gradients. Atheromas should be concentric or eccentric and irregular. The signal intensity of arteriosclerotic plaques is complex, depending on composition [28]. Arterial dissections produce a peripheral crescent, spiral, tram track, ring, or line of nonuniform width of typically bright signal along the walls of the involved vessel owing to clotted blood within the false lumen [29]. Intraluminal clotted blood would lie within the vessel in question and would have complex signal intensity depending on the age of the clot, the field strength of the MR imaging system, and the pulse sequence parameters chosen [29–34]. Intravascular clotted blood is most easily recognized as increased signal intensity on short TR/short TE– and long TR/long TE–weighted images [33, 34]. None of these conditions would demonstrate even-echo rephasing or flow-related enhancement. The reproducible pattern of the spatial misregistration artifact distinguishes it from other MR artifacts, particularly the chemical-shift artifact, which occurs only in the direction of the frequency-encoding gradient at the interface between tissues of different resonant frequencies.

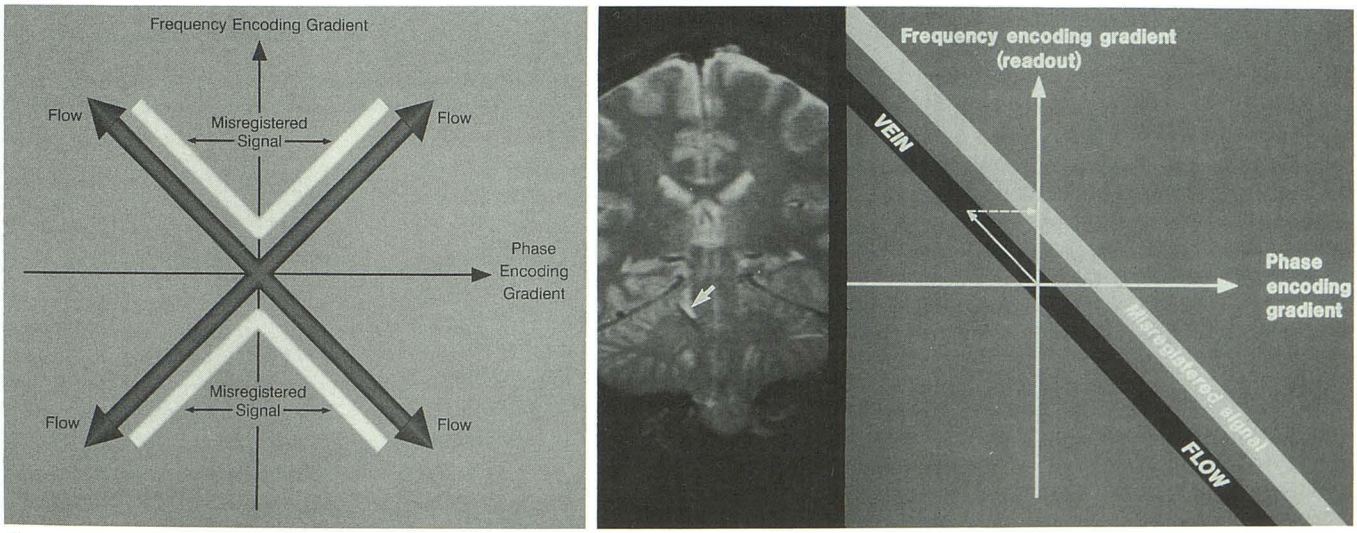

Fig. 11.—*A*, Schematic diagram used to predict direction of flow within vascular structure knowing orientation of phase- and frequency-encoding gradient axes and relationship of misregistered signal to parent vessel.

***B*, Second echo of T2-weighted coronal image (2000/35,70) and accompanying diagram of vessel and misregistered signal in relation to phase- and frequency-encoding gradients. Venous angioma drains peripherally toward tentorium, as would be predicted by displaced spatially misregistered signal (*arrow*).**

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