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Nuclear magnetic resonance (NMR) imaging with spin-echo techniques defines vascular structures with superb anatomic detail. Contrast agents are not necessary as there is intrinsic contrast between flowing blood and the vascular wall. The signal intensity from blood within the vessel lumen varies with the sequence of gradient and radiofrequency pulses used to generate the image as well as with the velocity of blood flow. Appropriate imaging techniques can optimize anatomic detail, distinguish slow from rapidly flowing blood, and serve to identify marked impairment or complete obstruction of flow in an artery or vein. Some examples of these principles in the intracranial circulation are illustrated.

Recent clinical trials of nuclear magnetic resonance (NMR) imagers have demonstrated their ability to generate images with superb contrast and spatial detail in multiple planes. Blood vessels are especially well defined, even in the absence of any contrast agents. The high contrast difference between flowing blood and the vascular wall provides a means of assessing vascular patency and luminal irregularity. The signal intensity, as well as contrast and spatial detail, vary considerably with the instrument and technique because of complex relations among the radiofrequency pulse sequence interval and echo delay, the rate of blood flow, and the presence or absence of turbulence. Our review elaborates and illustrates these principles.

Technical Considerations

Our NMR imager is a 3.5 kG superconducting magnet with saddle-shaped imaging coils having a 25 cm aperture for the head and a 55 cm aperture for the body. The gradients modifying the main magnetic field are 1 G/cm, with a rise time of 1 msec. Shielding from external radiofrequency sources is provided by enclosing the magnet and patient bed in cooper mesh. The imaging matrix is 128 × 256 resulting in a pixel size of 1.7 mm for field sizes up to 21 cm, a pixel size of 2.1 mm for field sizes up to 26 cm, and a pixel size of 2.5 mm for field sizes up to 32 cm. The sections are 7 mm thick. The distance between the centers of two contiguous sections is 11 mm.

The technique parameters for our spin-echo images are echo delays of 28 msec (first echo) and 56 msec (second echo) and pulse sequence intervals of 0.5, 1.0, and 1.5 sec. The echo delay (a) is the time between radiofrequency excitation of the nuclei and receipt of a signal or pulse echo from the nuclei. The pulse sequence interval (b) is the time between repeated radiofrequency perturbations of a volume in the sample.* Both the first and second echoes are measured in each pulse sequence interval. The second echo, at 56 msec, does not increase the overall imaging time significantly, because it only requires an additional 28 msec, which is considerably less than the pulse sequence interval, b. But it will affect the number of sections that can be obtained simultaneously. Multisection techniques involve imaging of additional sections during the pulse sequence intervals; the number of additional sections that can be imaged depends on the length of the pulse sequence interval. The overall imaging time is the product of: (1) the pulse sequence interval length, (2) the number of lines along the y axis, and (3) the number of times the

* Editor's note.—Echo time (a) and repetition time (b) are identified as TE and TR, respectively, by other authors in this issue.

TABLE 1: **NMR Spin-Echo Techniques at the University of California, San Francisco**

Pulse Sequence Interval (sec)	Echo Delay (msec)	No. of Sections	Imaging Time (min)
0.5	28, 56	5	4.3
1.0	28, 56	5	8.5
1.5	28, 56	15	6.5, 13.0*

* Two and four averages, respectively.

signal is averaged. The greater the number of averages, the greater the signal-to-noise ratio. Our imager uses the sequences listed in table 1.

For spin-echo images, the intensity I is calculated by the equation: $I = HF(v)\exp(-a/T_2)[1 - \exp(-b/T_1)]$. H is the local hydrogen density, f(v) describes the effect on intensity from the speed with which hydrogen nuclei move through the imaged region and the fraction of moving nuclei; a is the echo delay; b is the pulse sequence interval; and T_1 and T_2 are relaxation times. The term f(v) indicates that blood flow will change signal intensity. Note that a and b are temporal variables of this instrument and will affect the image intensity even if Hf(v) is constant. Inversion-recovery images are obtained with echo delays of 28 and 56 msec and a pulse sequence interval of 1 sec. The recovery time is 0.42 sec. Five adjacent sections are imaged in 8.5 min. The basic principles of NMR and their impact on the information content and quality of the resulting images are fully discussed in two recent reports and will not be addressed further [1, 2].

Effects of Motion on the NMR Signal

The intensity of the NMR signal is influenced both by T_1 and T_2 relaxation times as well as the motion of the hydrogen nuclei being imaged. For example, if the hydrogen nuclei pass through the imaged volume faster than the time it takes to perform an imaging sequence, the signal will be zero. The signal intensity from flowing blood is a function of the percentage of moving hydrogen nuclei, their velocity, and the temporal parameters of the imaging technique. Moving nuclei would be expected to have very small or absent signals because they do not experience the same radiofrequency and gradient pulse sequence as stationary protons. However, the signal intensity emitted by the protons in blood is a more complex behavior [2].

To appreciate the effects of high velocity blood on the NMR signal intensity, an understanding of stationary and slowly flowing blood is necessary. The protons in stationary blood in an occluded vessel, as well as extravascular protons, yield an NMR signal intensity based on the degree of magnetization they achieve between repeated radiofrequency perturbation (fig. 1). Let us arbitrarily define the signal intensity available from stationary protons as 100%, that is, the signal intensity is always considered as 100%, regardless of the percentage of remagnetization the immobile protons attain during the pulse sequence interval.

Blood flowing at low velocities produces a signal of greater intensity than stationary blood (fig. 2). The increased signal is due to the introduction of fully magnetized protons into the image plane. These previously unperturbed protons yield more signal when first excited than the partly demagnetized protons that remain within the imaged section. The
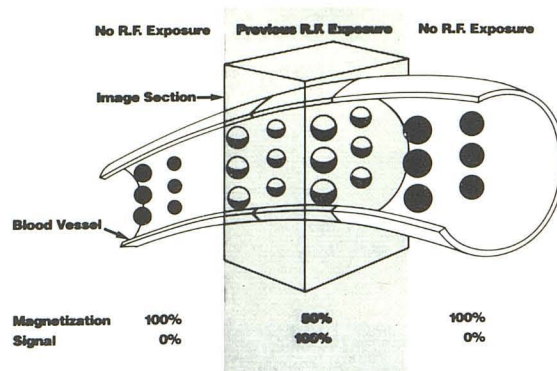

Fig. 1.—NMR signal intensity of stationary blood. *Circles* represent protons. Their degree of magnetization is indicated by proportional volume in *black*. Protons outside imaged section do not experience radiofrequency (R.F.) excitation and remain fully magnetized. These protons emit no radiofrequency signal as they are not perturbed by radiofrequency pulses. Protons in stationary blood (as well as extravascular protons) within image plane experience repetitive radiofrequency excitation, preventing restoration of full equilibrium magnetization, and are represented by *partly filled circles*. In this simulation, 50% magnetization is assumed for these protons. Such stationary, partly magnetized protons are arbitrarily considered to have 100% signal intensity in this example.
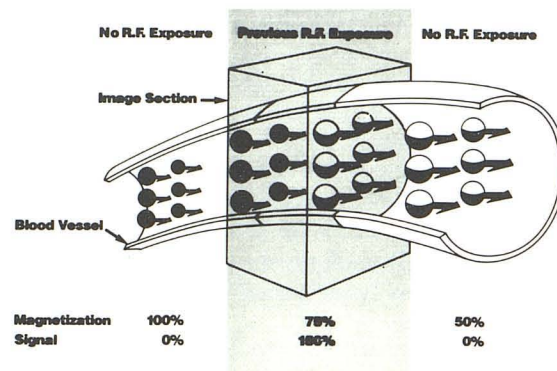

Fig. 2.—Paradoxic enhancement: slow flow. At low flow velocities, paradoxic enhancement is seen. Slowly flowing protons, before entering imaged section, have not experienced radiofrequency (R.F.) excitation and have full magnetization (*black*). Within imaged plane, two groups of protons are now evident. Protons just entering image section have 100% magnetization (*black circles*) and will yield more signal when first perturbed. Protons previously present in image section have only partial magnetization, having been previously perturbed (*half-filled circles*). Assuming an equal number of protons with full and half magnetization, a population with an average magnetization of 75% can be said to be present. If, by our convention, stationary protons with 50% magnetization gave 100% signal intensity (fig. 1), the sample of protons with a magnetization of 75% will yield an intensity of 150% relative to former.

result is an ensemble of protons, composed of those not previously irradiated and having full magnetization, flowing slowly enough to fully interact with the imaging sequence, and those previously irradiated and having partial magnetization. The NMR intensity of the ensemble will, consequently, exceed 100%. This effect has been termed *paradoxic enhancement* [2].

Nuclei moving at a comparatively high velocity will escape the imaged volume during the interval required for radiofrequency and gradient pulse administration (fig. 3). Conse-
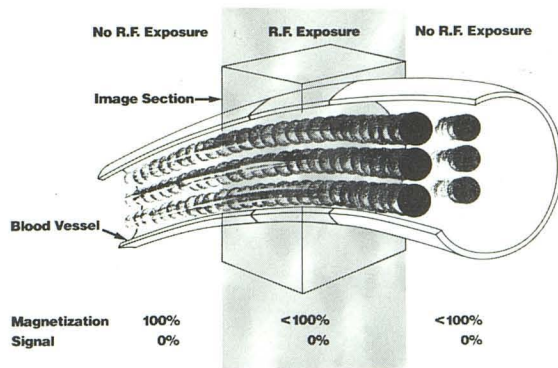
Fig. 3.—NMR intensity of high-velocity blood. Protons are within image section for a short time and do not experience a complete radiofrequency (R.F.) and gradient pulse sequence. Consequently, although they may be partly demagnetized, there will essentially be no observable radiofrequency signal emitted.

Fig. 5.—Internal carotid artery, sagittal spin-echo image. Internal carotid artery (black arrows) has no signal when a = 28 msec and increases in intensity in transverse part when a = 56 msec. Increased intensity is most likely secondary to turbulent flow. Cavernous sinus (white arrows) has higher intensity because of slower blood flow.

Fig. 4.—Normal vessels, transverse spin-echo image. Internal carotid (black arrows) and basilar arteries (arrowhead) and right transverse sinus have no signal. Boundary layer flow is in left sigmoid sinus (white arrow).

quently, these protons produce no detectable signal. For example, let us consider an imaging sequence of 50 msec for a 5 mm section. Moving nuclei will traverse this distance in 50 msec if their velocity is 10 cm/sec. For this case, the threshold velocity above which moving nuclei do not contribute signal is about 10 cm/sec.

Illustrative Examples

During a 6 month period, a total of 75 subjects underwent cranial NMR examination. Sixty-eight were patients and seven were normal volunteers. They were 2 months to 80 years old. The clinical diagnoses included degenerative brain disease, primary intracranial tumors, metastases, inflammatory disease, ischemic lesions, arteriovenous malformations, trauma, and various neurologic abnormalities.
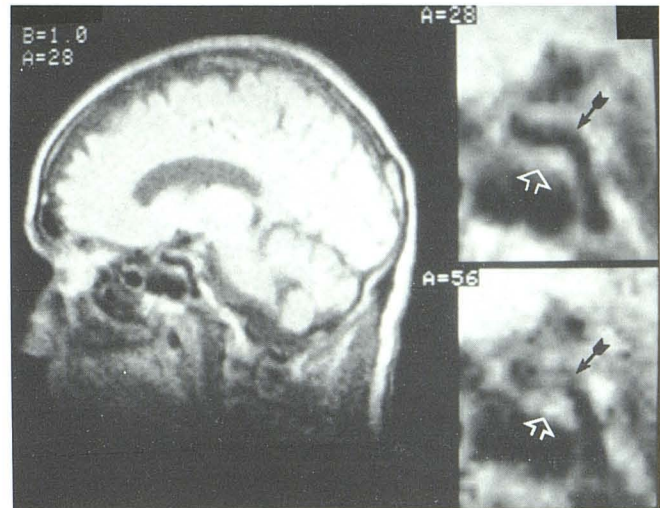
Their NMR images were reviewed to determine those situations in which normal vessels were most clearly identified and those imaging sequences, flow patterns, or pathologic processes in which the NMR signal was altered.

Normal Vessels

Vascular structures are identified by NMR in the transverse, sagittal, and coronal planes (figs. 4–6). Multiplanar imaging is advantageous, because, depending on the orientation of the vessel, the entire cross section will be identified in one of the three planes. Images perpendicular to the long axis of a vessel are devoid of the effects due to partial-volume averaging. On the other hand, sagittal sections through the neck may image only a part of the course of carotid artery or jugular vein (fig. 7). The interpretation of pathologic atherosclerotic narrowing may be problematic unless the vessel is completely within the imaged section.

Variation with Technique

Variations of technique parameters will affect the signal intensity of flowing blood. With our technique, images obtained with a single pulse sequence interval and two echo delays for a given velocity have shown a higher intensity signal for blood in the second echo compared with the first (figs. 8 and 9).

Variation with Blood Flow Velocity

NMR imaging techniques detect normal vascular structures and also demonstrate the abnormal flow seen in arteriovenous malformations. The signal intensity varies from the low intensity in the fast moving blood of the arteries
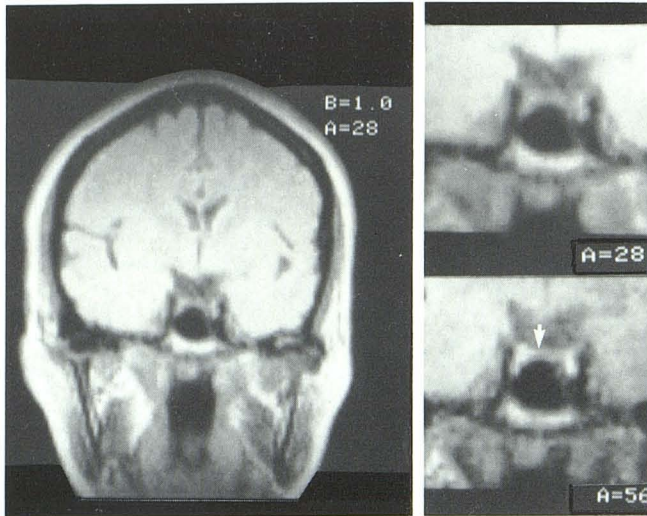
Fig. 6.—Internal carotid arteries and cavernous sinus, coronal spin-echo image. Internal carotid arteries have no signal. Increased signal in cavernous sinus and either pituitary gland or circular sinus with a = 56 msec (*arrow*).
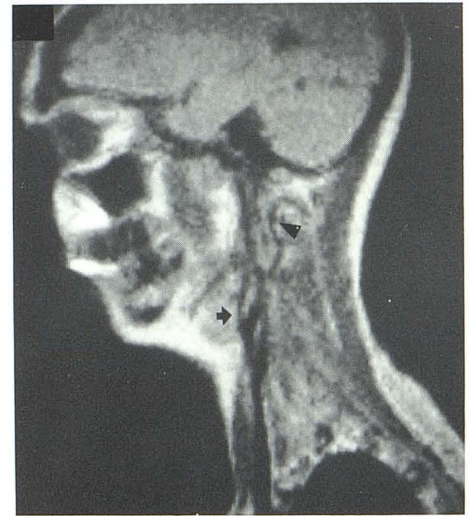

Fig. 7.—Cervical carotid arteries, sagittal spin-echo image. Parts of common carotid artery and its bifurcation into internal and external carotid arteries (*arrow*), vertebral artery (*arrowhead*), and internal jugular vein are seen as low intensity.

supplying and veins draining the malformation to the high intensity of the matrix where blood flow is slower (fig. 10).

Slow Flow versus Occlusion

The changes in signal intensity with different technical parameters are useful in recognizing pathologic alterations in blood flow patterns. Distinguishing slow flow from stationary blood depends on the signal intensity variations between the first and the second echo delays. Vascular occlusion will result in a high-intensity signal that diminishes somewhat between the first and second echo delays (fig. 11). If there is not complete occlusion, and slow flow is present, the signal intensity will be equal or higher for an echo delay of 56 msec than for 28 msec (fig. 12).

Location of Section

The signal intensity of flowing blood is affected not only by the technical parameters, but also by the position of the section in the multisection sequence. The first section of a sequence demonstrates the highest intensity for slowly flowing blood, and the signal diminishes in subsequent images (fig. 13). This then gives us an indicator of direction of flow, which aids in distinguishing between arteries and veins.

Discussion

NMR imaging depicts vasculature in exquisite antomic detail [2–9]. The high contrast between normal flowing blood and the vascular wall defines precisely the luminal contours. This has application in carotid atherosclerosis, which is responsible for significant morbidity and mortality. NMR of human autopsy specimens has demonstrated atherosclerotic plaques clearly [9]. Plaques are imaged because
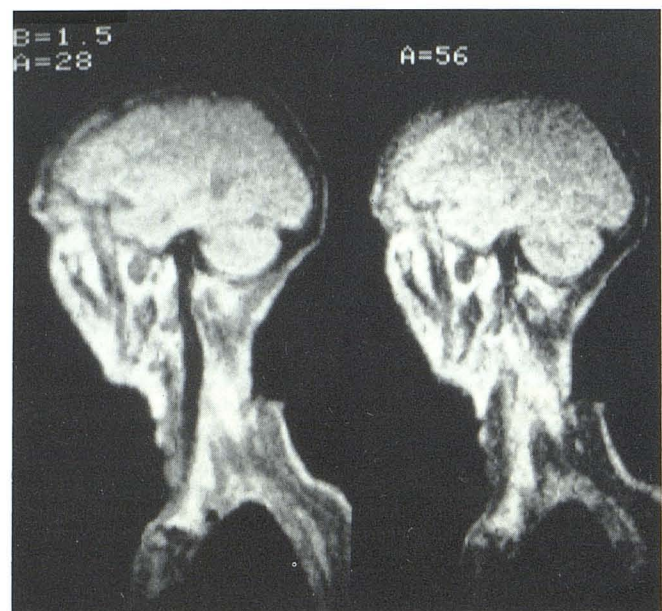

Fig. 8.—Internal jugular vein, sagittal spin-echo image. Internal jugular vein is sharply delineated when a = 28 msec. However, with a = 56 msec, contrast between flowing blood and perivascular tissues decreases.

of their intraluminal location, resulting in a positive intensity signal where there would normally be no signal. Also, the characteristic lipid component of arteriosclerotic deposits produces a high-intensity signal owing to the short T_1 and long T_2 relaxation times.

Variations in the signal intensity of flowing blood are observed secondary to the technique parameters or the section position. With our technique, images with two echo delays will show, for any one velocity, a higher-intensity
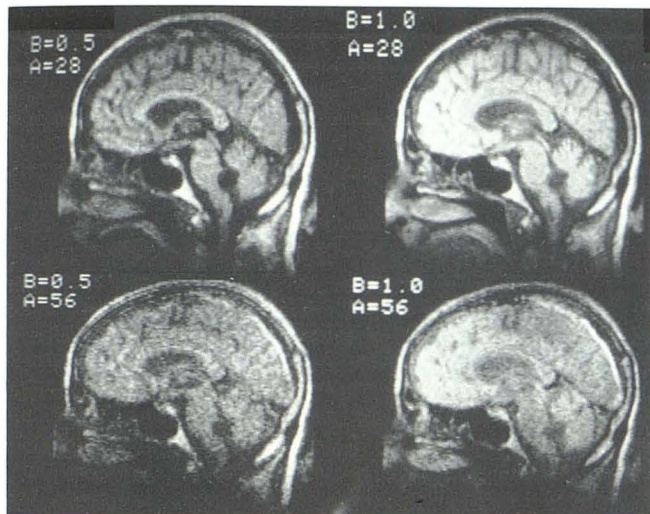
Fig. 9.—Superior sagittal sinus, sagittal spin-echo image. Superior sagittal sinus is black and inseparable from inner table of skull when a = 28 msec. Signal intensity of sinus and cerebrospinal fluid increases in second echo image.

Fig. 11.—Occluded left carotid artery, transverse spin-echo image. Normal right internal jugular vein (*small closed arrow*) and internal (*large closed arrow*) and external (*open arrow*) carotid arteries with low signal. Occlusion of left carotid artery is evident, as no vessel is distinguishable from perivascular structures.
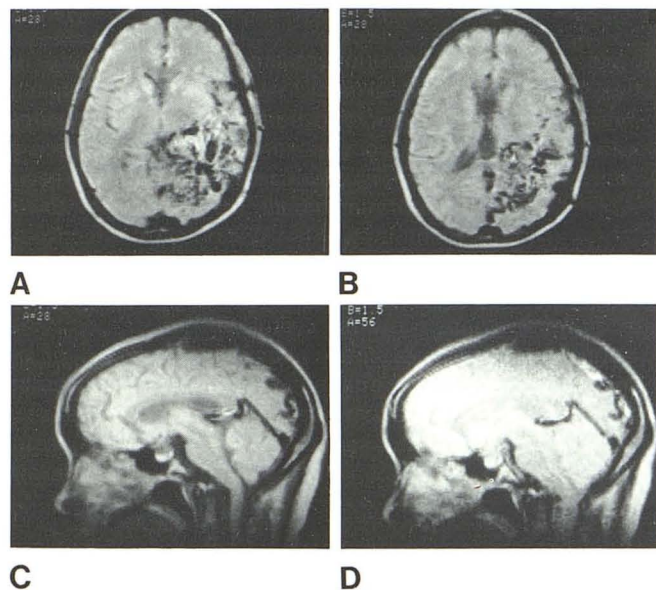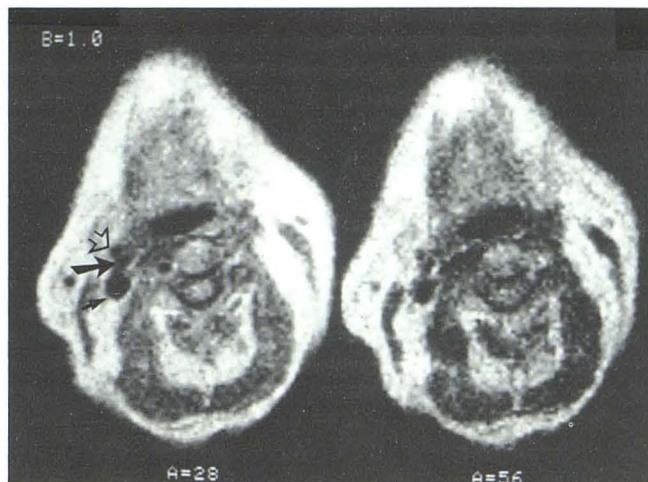

A   B

C   D

Fig. 10.—Parietal lobe arteriovenous malformation. **A** and **B**, transverse spin-echo images. Prominent low-intensity vasculature results from rapid flow. **C** and **D**, Sagittal spin-echo images. Prominent cortical veins and straight sinus caused by rapid flow from AVM. Basilar artery is visualized on second echo image.
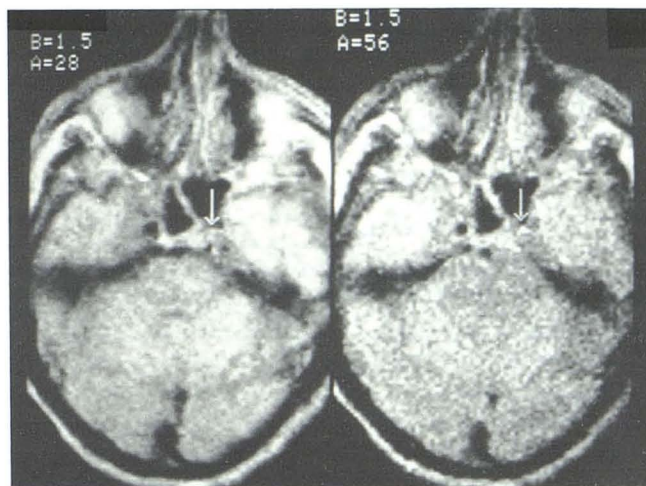

Fig. 12.—Normal right internal carotid and basilar arteries, transverse spin-echo images. Slow flow in left internal carotid artery (*arrows*) with paradoxic enhancement.

signal for the second echo delay. If the pulse sequence interval is decreased, the stationary or background nuclei lose a larger proportion of signal compared with the moving nuclei because they recover less magnetization between each pulse, while flow adds fully magnetized protons to the sample. Accordingly, paradoxic enhancement will be increased for short pulse sequence intervals.

The multisection technique, which decreases the total imaging time, modifies the signal intensity of slowly flowing

protons [1]. The enhancement of the NMR signal is greatest in the sections first entered by the blood. These sections receive protons that have not experienced a prior radiofrequency or gradient pulse excitation and will, therefore, provide maximal signal. The signal in the subsequent sections is decreased or absent, because the protons were perturbed in the previous section. A multisection sequence oriented with the most cephalad section first and the most caudad section last will demonstrate enhancement of arterial blood in the last caudad section for neck images and in the first cephalad section for body images. Venous blood will show the opposite pattern, with enhancement in the first cephalad section for neck images and in the last caudad
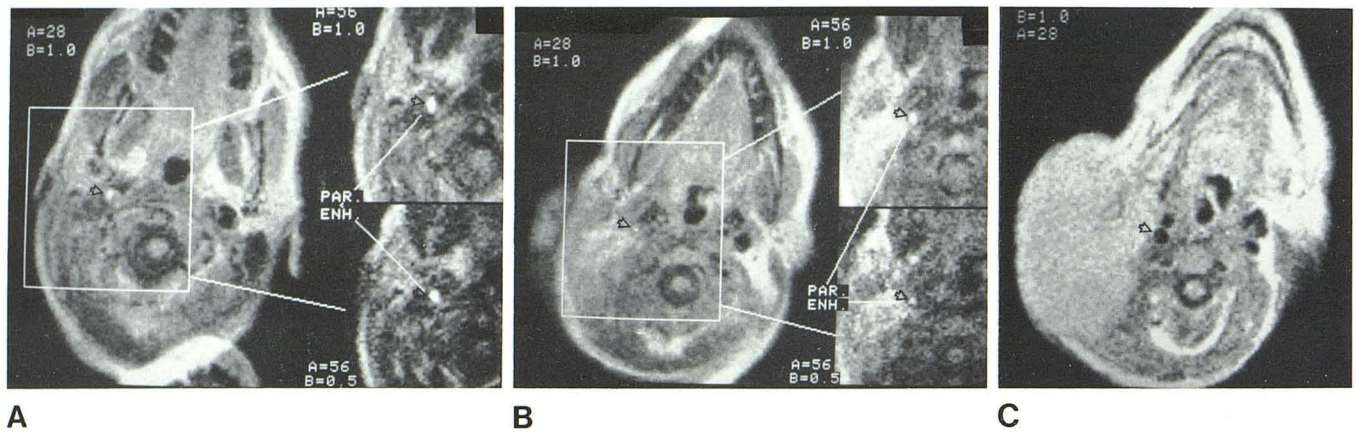
Fig. 13.—Squamous cell carinoma of right neck producing partial obstruction of right internal jugular vein (*arrows*). Transverse spin-echo images in sequential sections. **A**, High-intensity signal in right internal jugular vein indicates paradoxic enhancement. **B**, Intensity decreases. **C**, No evidence of enhancement in black internal jugular vein.
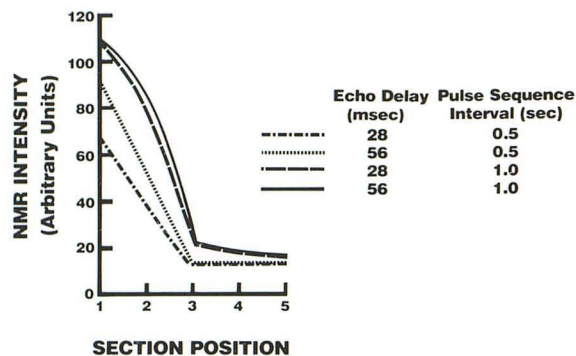

Fig. 14.—NMR intensity of right internal jugular vein from fig. 13 as function of section position in multisection sequence. Intensity of internal jugular vein is highest in first craniad section of sequence and decreases in subsequent caudad images. Intensity variation reflects directionality of flow and allows differentiation of arterial and venous structures.

section for body images. These enhancement patterns reflect the opposing directions of flow of arterial and venous blood (figs. 13 and 14). The volume irradiated also affects flow phenomena. If the excited volume is large compared with the sections being imaged, stationary and flowing blood will be represented with a similar intensity, as nuclei outside and inside the imaged volume undergo the same sequence of perturbation [3].

Another NMR flow phenomenon is the ring, or boundary layer pattern. The higher-velocity blood is centrally located and exhibits no discernible NMR signal. Slowly flowing blood near the wall interacts with the radiofrequency and gradient pulse sequence and has definable signal [9]. Turbulence will also alter the signal intensity of flowing blood. The effect of turbulence is to maintain a subpopulation of protons within the imaged plane and, thus, increase the signal.

The demonstration of vascular structures and their intensity variations secondary to manipulation of the technique parameters have raised the possibility of quantifying flow with NMR. On the basis of the bolus technique or flow

graphs of Singer [10, 11], experimental models can be used conceptually to quantitate blood flow. However, the full potential of NMR in the evaluation of the hemodynamics in normal and pathologic conditions will require much further investigation.

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