Diagnostic Radiology

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DEDICATION

"Each honest calling, each walk of life, has its own elite, its own aristocracy, based on excellence of performance."

JAMES BRYANT CONANT
Harvard Baccalaureate Sermon
June 16, 1940

This book is dedicated to the physicians who are residents and fellows in Radiology at the University of California, San Francisco.

Preface

The Department of Radiology of the University of California Medical Center in San Francisco is proud to present this synopsis of the state of the art in the field of radiology. Because of the increasing importance of magnetic resonance imaging and interventional techniques, as well as computed tomography and ultrasound, the challenge for us was to select stimulating topics and authorities who could offer both expert commentary on the techniques involved in these new procedures and critical commentary on their effectiveness. The topics presented are pragmatically oriented; they have been chosen with the intent of aiding clinical radiologists in their medical practice.

The editors thank Mrs. Renee Sauers and the Postgraduate Education Section of the Department of Radiology of the University of California, San Francisco, for all the time and effort that they have contributed to ensure the success of this book. We extend our thanks to Antonio Padial for editing chapters, to Charlie Scribner and the University of California Printing Services, whose expertise and cooperation made this publication possible, and to Anne Poirier for administrative assistance. We would also like to express our sincere thanks to those who contributed the chapters.

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Technical Advances in MRI

Leon Kaufman, Ph.D.

Advances in magnetic resonance imaging (MRI) capabilities can arise through changes in hardware or software. Software consists of three major components: acquisition software that affects the object contrast obtained in the image; acquisition software that determines spatial resolution, imaging time, and signal-to-noise (S/N) levels; and display and data processing software. The hardware can be thought of as the framework that supports the software, with the latter providing the motive power for the systems.

Some believe that a high field magnet is the key to improving operation of the MR imager. To achieve an improvement in performance, the operating field would be raised. If, however, benefits accrue from high field operation, it would seem advantageous to operate at a high field from the start. Over the last 2 years, accumulated independent experience with high field operation has demonstrated that high field magnets do not lead to improved performance.[1,2] Over the last 2 years we have achieved a threefold increase in S/N levels at 3.5 KGauss by changing the circuitry and shape of RF coils and by improving some components of the signal detection electronics. Coupled with increased efficiency of data acquisition introduced by software changes, this improvement has meant the quadrupling of S/N levels in 2 years.

S/N levels are a useful technological indicator; however, from a diagnostic point of view, S/N levels for any one imaging procedure have only a discrete effect: Below a certain S/N level, the diagnosis is uncertain; above the level where the needed confidence is reached, further increases do not improve diagnostic accuracy.

Impressive as increases in S/N levels are, two software developments have had significantly greater clinical impact. The first has been the introduction of careful sequence tuning methods that lead to single-pass, 2-D FT contiguous sections without gaps.[3] With these sequences it is possible to measure the NMR parameters T_1, T_2, and N(H), whose importance is discussed below. Together with these changes, 2-D FT sections as thin as 2.5 mm are obtained (Fig. 1). All of the imaging is performed with either one acquisition (no averaging of data) (Fig. 2) or two acquisitions (one average of data), obtaining from one to four echoes, depending on need. Low acquisition numbers are made possible not only by intrinsic imager S/N levels that do not require noise averaging but also by acquisition sequences that avoid serious artifacts otherwise cancelled by the averaging process. It is now possible to image with a sequence that offers high sensitivity to brain disease (TR = 2 seconds at 3.5 KGauss) and obtain 20 sections with two echoes and a 1.7 mm resolution in 4.3 minutes. A second acquisition for obtaining enough data to compute T_1, T_2, and N(H) images requires another 1.1 minutes for 10 sections or 2.2 minutes for 20. These data can be acquired and be readied for viewing in 15 minutes, including patient set up time as well as tuning and sequence changeover time, a throughput that can be hardly matched by an x-ray computed tomography (CT) scanner. Ultra-high resolution (0.5 mm) images in sections 5 mm thick can be obtained with single-average imaging as well (Fig. 3). Conceptually, neither S/N nor imaging time have reached operational plateaus, with significant improvements (factors of 2–4) possible for each. Section thicknesses of 1 mm are potentially attainable as well.

Much has been written about the value of various spin echo (SE) and inversion recovery (IR) sequences in relation to the development of sequences that optimize object contrast. It is now well understood and demonstrated that long TR sequences of moderate TE values

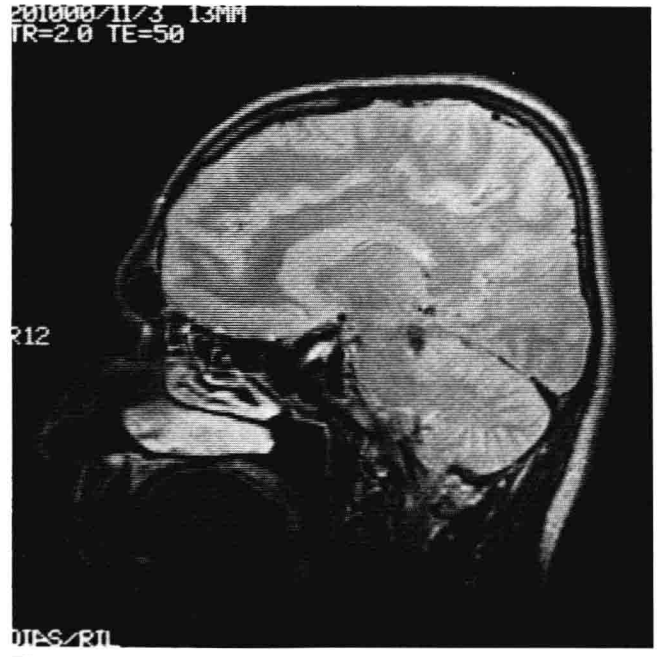

Figure 1. Sagittal section, 0.95 × 0.95 mm resolution, 2.5 mm thick, obtained at 3.5 KGauss.

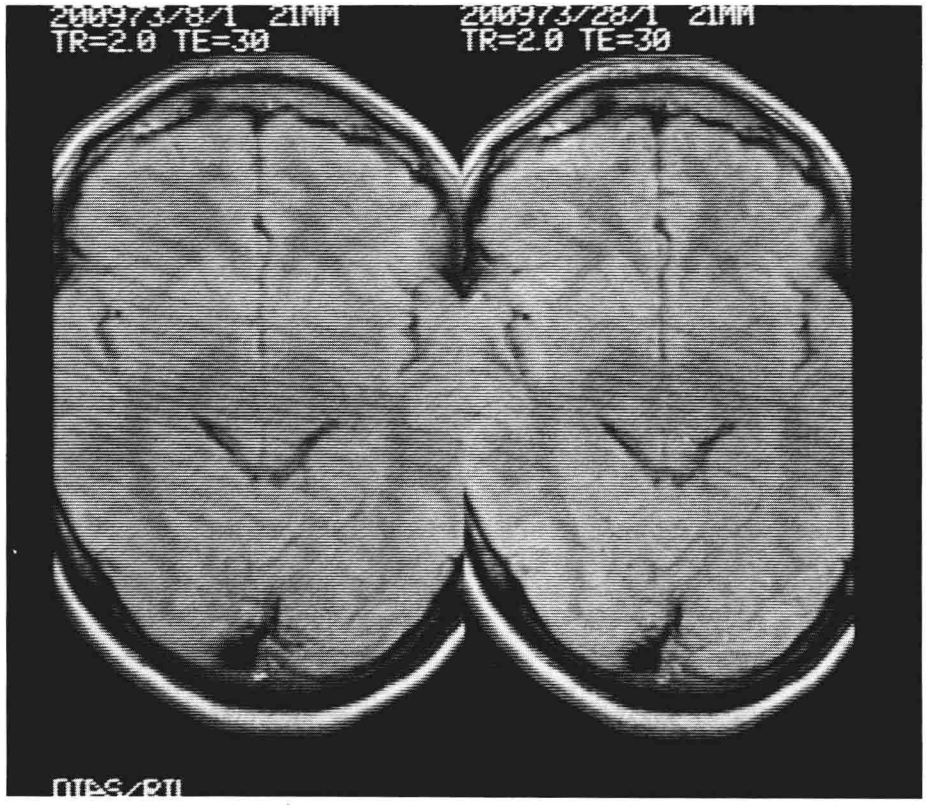

Figure 2. Comparison of a transaxial section obtained with two acquisitions (single-average) on the left and one acquisition (no averaging) on the right. Twenty sections with two echoes are obtained in 8.5 mm for 0.95 mm resolution, and in 4.3 mm for 1.7 mm resolution.

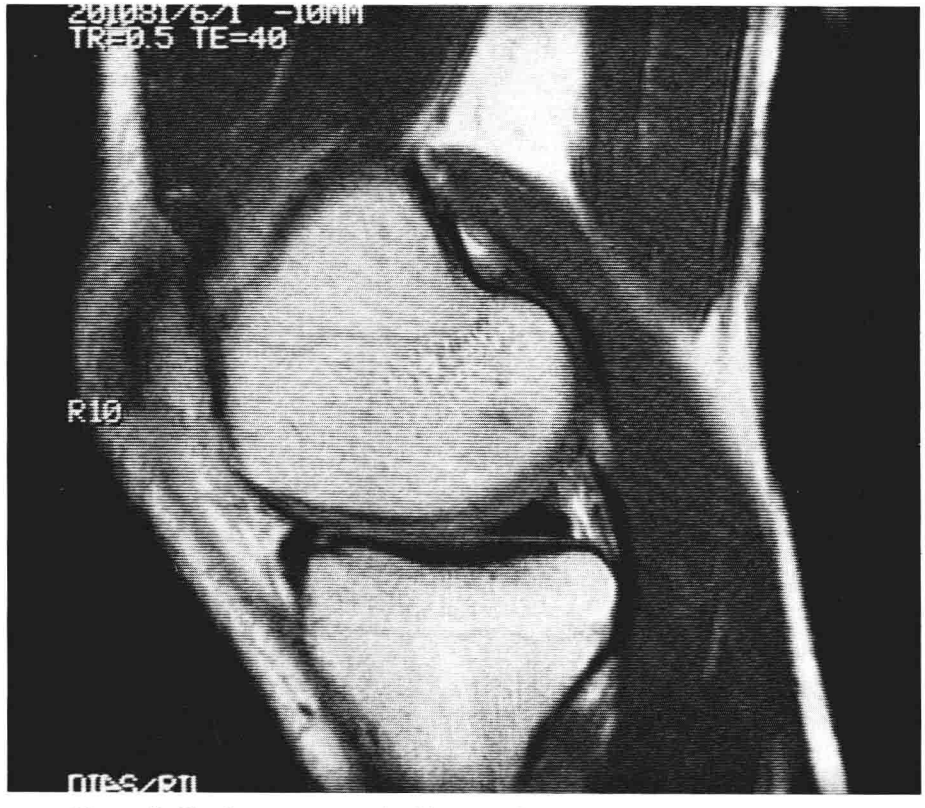

Figure 3. Single-average sagittal image of a knee. The spatial resolution is 0.5 × 0.5 mm and section thickness is 5 mm.

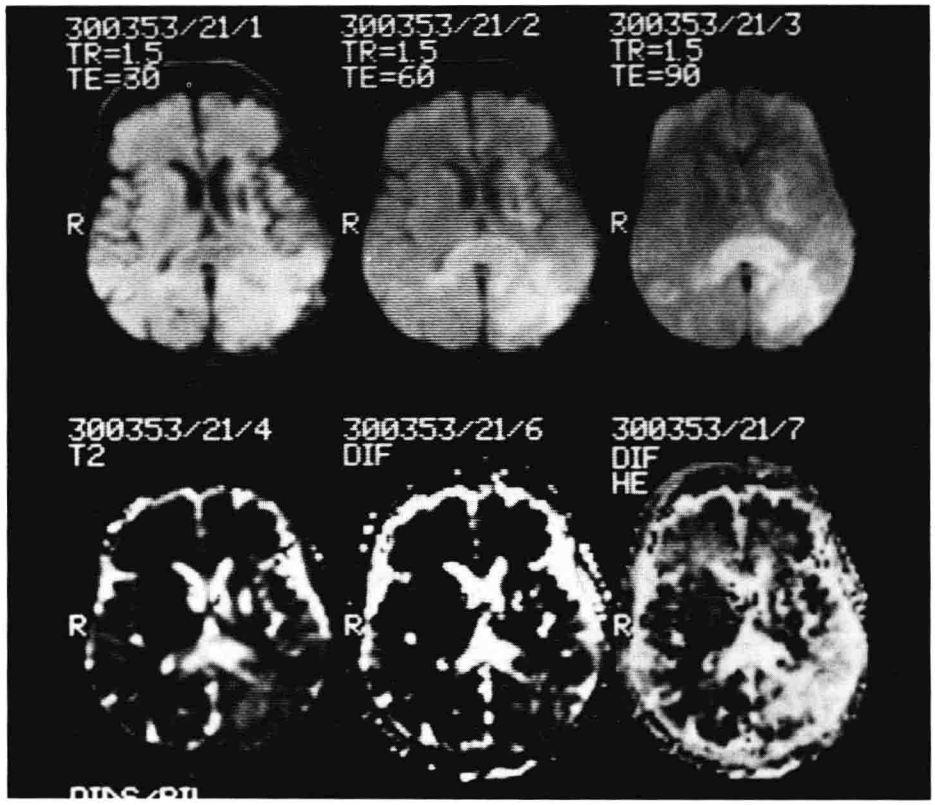

Figure 4. Top row shows multiecho images of a patient with a primary brain tumor after radiation therapy and a known, old, left basal ganglial infarction. At bottom left is a T_2 image calculated from the three echoes above. In the bottom center is a diffusion image shown in full dynamic range. Note that cerebrospinal fluid, the tumor, and infarct show large diffusion. At right is a histogram-equalized diffusion image, showing an area of edema on the side of the tumor.

($TR>2T$, $TE\sim T_2$) provide valuable general-purpose screening sequences. In some cases, late echoes ($TE\sim 2T_2$) and short TR sequences ($TR<T_1$) serve specific purposes, as do IR sequences with long TR values.[4,5]

An exciting development is the addition of sequences that provide data not available through SE or IR imaging. One of these is water-fat imaging—two acquisitions are used to provide two image sets differentiating these two substances.[6] An even less-explored technique generates images of molecular self-diffusion[7] (Fig. 4), an indication of the ability of a water molecule to travel, determined by its root mean square velocity and the size of the compartment holding the water.

Simply expanding the number of sequences used on a patient (by adding sequences and reducing the acquisition time for each) does not necessarily add to the diagnostic value of the procedure. It is difficult for an interpreter to integrate the data in ten images—four SE images; one IR sequence; one T_1, one T_2, and one N(H) image; one water-fat image; and one diffusion image—for just five selected sections of 1 patient. The interpreter can hardly be expected to integrate the data in five sections (ten images per section) of 15 patients daily. If full use is to be made of the information in the MR image, data integration techniques must be used. Such integration techniques reduce the information in the image set to just two or three images that carry information appropriate for the study. Although the awareness of the need for such processing is lacking in the user community, such awareness is necessary unless MRI is to end by being "read like CT."[8] The following are examples of what we call diagnosis-related processing.

If enough data have been generated in the imaging process (at minimum, a 1-echo SE with short TR and a 2-echo, long-TR SE sequence), then T_1, T_2, and N(H) images can be calculated. Given two areas of interest, new SE or IR images that maximize signal difference/noise levels between the two regions can be generated (Fig. 5). Three or more regions of interest can be considered in selecting the imaging parameters for the calculated image.[4] Late echo images can be calculated from double-echo procedures (Fig. 6). These carry diagnostic information equivalent to multiple echoes at fixed echo times and obviate the need for acquiring, processing, and archiving these multiple echoes.[9]

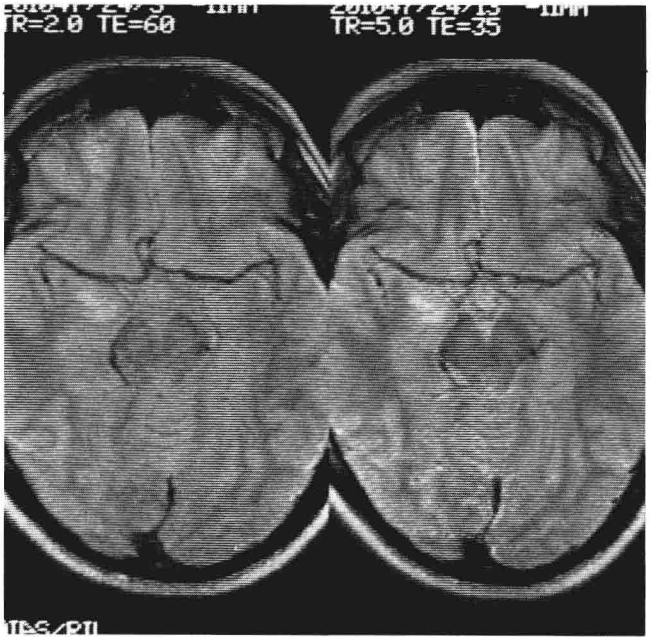

Figure 5. Transverse image of patient with a temporal lobe tumor. The sequences used for acquisition had TR = 0.5 and 2 seconds, and TE = 30 and 60 msec. A computer-based analysis predicts higher object contrast for a TR = 5 second-sequence, which would require twice the acquisition time. Instead, a TR = 5 seconds, TE = 35 msec image was computed. Compared to the acquired image that best showed the tumor (left), the computed image (right) has higher contrast and better S/N.

These calculated images also obviate the need to obtain a large set of SE and IR sequences in the hope of "covering the field." If two sequences with short and long TR are performed, the TR, TE (for SE) and TR, TE, TI (for IR) space can be sampled under the guidance of a computer. This process can be automatic once the regions of interest are identified. It is worth noting that this processing is useful only if accurate T_1, T_2, and N(H) data are generated in multislice imaging. Obtaining these "basic" images, even if an observer were never to look at them, is necessary to allow for calculated images.[10] The images that are calculated need not necessarily reproduce those that can be acquired. For instance, because in some diseases tissue water increases result in lengthening of T_1 and T_2 and an elevation of N(H), a "liquefaction" image can be produced (Fig. 7) by multiplicatively weighting these three parameters.[4] Probably the most speculative effort in data integration involves finding pixels with common characteristic NMR parameters. These pixels can then be grouped and assigned a single identifying mark (gray scale or color). Reasonable tissue type images have been obtained (Fig. 8) with fully automated programs, and further improvements are possible.[11]

Although the previously described developments are applicable in a general way, some programs are being developed for specific applications. An area where considerable success is being achieved is flow identification and imaging. One often-encountered problem

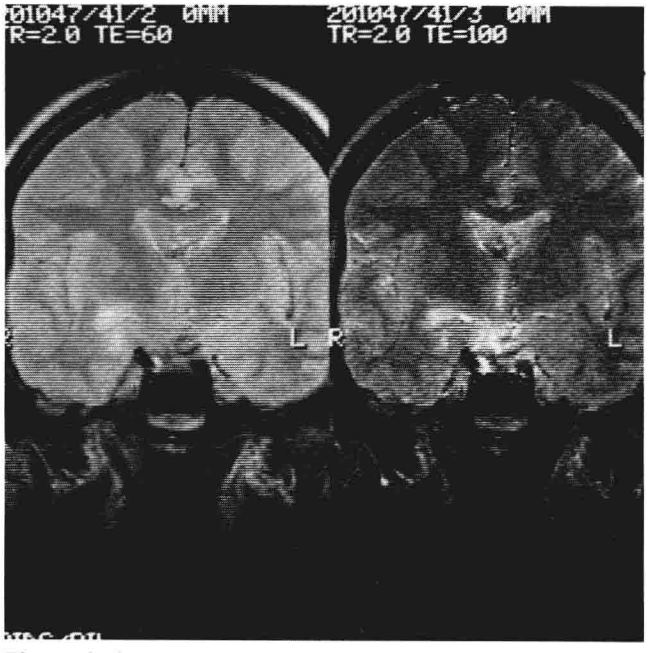

Figure 6. Coronal image of the same patient with a temporal lobe tumor. The acquisition included 30 and 60 msec echoes (left). From these a 100 msec echo was computed (right).

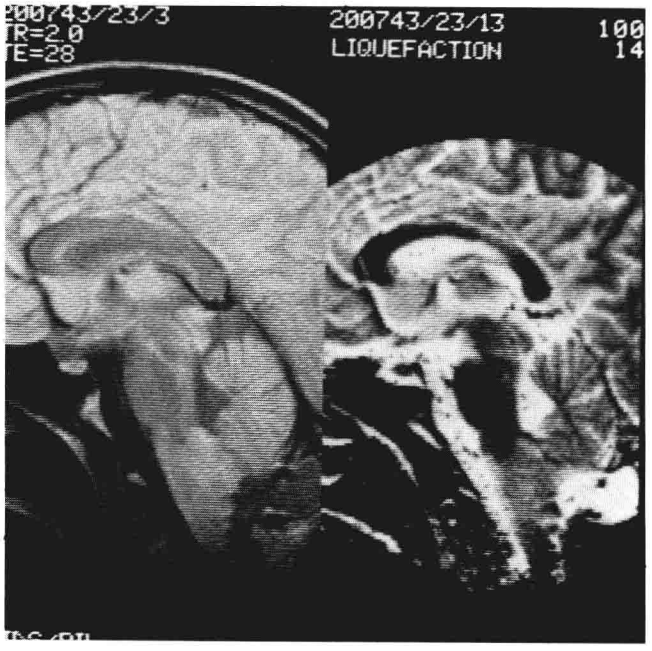

Figure 7. Acquired sagittal spin echo image (left) with a long TR shows mass effect due to a tumor in the medulla. Signal intensity alone does not reliably identify the presence of an abnormality. The liquefaction image at right shows significantly higher signal intensity in the tumor as compared to normal white matter.

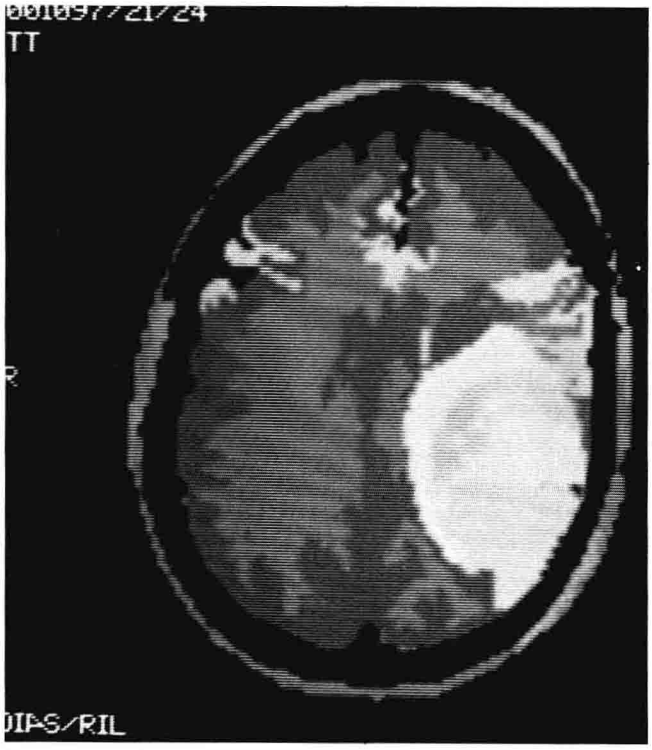

Figure 8. Tissue type image of a patient with a parietal tumor. The algorithm identifies cerebrospinal fluid, gray matter, white matter, and four compartments within the lesion.

is the patent lumen that produces signals suggesting slow flow or turbulence. Under these conditions, it is difficult to assess the degree of obstruction in the vessel. In all but the most extreme cases, the signal-producing regions of the lumen show relatively high signal in the second echo of a double-echo procedure. It is difficult to locate these regions by viewing the two echoes, but a computer program can search these regions and paint them with zero or very high intensity in the image, thus making them appear as expected where flow exists.

The tomographic format of MRI permits exquisite delineation of the lumen of major vessels and identification of lesions. Nevertheless, it is difficult to visualize vessel patterns from these images. Using flow identification signatures (low intensity or elevated second echo signal), observers can locate vessels in three dimensions.[12] Vessels can then be redisplayed from any viewing direction. The width of the displayed vessel can be the cross section from the viewing direction, or width can be renormalized to represent the true patent area (i.e., the vessel is circularized, and its diameter determines width in the image) (Fig. 9). Color can be used to code depth, patent area, or qualitative flow. This latter mode strikingly demonstrates differential flow characteristics in aortic dissection and turbulence around lesions.[13]

The advances described here do not represent the limits of what is conceptually achievable. As we gain

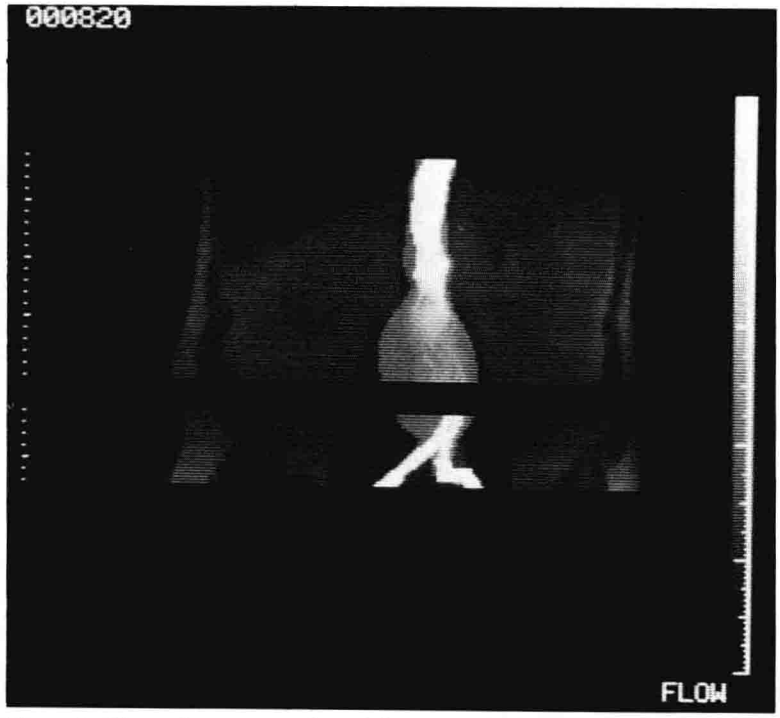

Figure 9. Coronal reconstruction of the aorta, obtained from two transverse acquisitions. (The gap in the reconstruction is from the displacement between acquisitions.) Plaque is shown in gray. In the lumen, high intensity represents faster flow. Flow is disturbed where the lumen undergoes widening and a bend. Turbulent patterns are evident in the region of the aneurysm.