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پروپوزال نویسی
Magnetic Resonance Angiography: Physical Principles and Clinical Applications

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Introduction

For many decades, radiologic evaluation of vascular disease has been with contrast-enhanced x-ray angiography. This requires catheterization and the use of iodinated contrast, therefore, excluding patients with renal insufficiency and those who are severely allergic to the contrast. The study is often performed in a single plane and, frequently, there is poor visualization of vessels distal to severe occlusive disease due to contrast dilution effects.[1] Over the past few years, Magnetic Resonance Angiography (MRA) has emerged as an accurate modality among the available noninvasive methods for vascular imaging.[2] Three dimensional data acquisition with projections along multiple orientations and with quantitative flow information has improved our ability to accurately diagnose vascular disease. The high spatial, high temporal resolution and patient acceptability has led many to further investigate the role of MR imaging in vascular disease. Based on the early successes in carotid and intracranial MR angiography, new interest has emerged in imaging the abdominal and peripheral vessels.[3][4] Furthermore, an accurate display of the entire body vasculature has become feasible with the introduction of stepping table technology.

Flow and magnetic resonance

The principle of observing vascular flow signal in MRI is based on exploiting the difference between signals from flowing blood and the signal from the surrounding stationary tissues. The magnetization of protons resulting in an image pixel depends on the variation in the intrinsic parameters ( T1, T2, velocity and spin density ) of flowing blood and the surrounding tissue protons as well as on the extrinsic parameters ( TR, TE, flip angle, slice thickness and orientation ) of the pulse sequences.[5] It is known that the greater the difference between the relaxation times of protons of blood and surrounding tissue, the greater is the contrast between the two. Therefore, one exploits relaxation time differences by using appropriate pulse sequence parameters to discriminate signal differences between blood and the surrounding tissue protons.

In conventional spin-echo MR imaging flow can occasionally be a source of artifacts that can impair our ability to analyze images. However, the flow effects can be used to our advantage to visualize anatomy.[6] The absence of signal, due to flow in a spin-echo, is called "signal void", and has been used to detect the presence of flow (Fig.1 A,B). However, these flow void imaging techniques are time consuming requiring cooperative patients and do not provide 3-dimensional display of vascular anatomy. The rapid evolution in vascular MR imaging began when the use of GRE techniques demonstrated
Fig. 1A: In spin-echo imaging, the stationary spins (protons) experience both 90° and 180° RF excitation pulses per phase encode, thus, provide signal. Spins, which are not stationary, on the other hand, will not experience both RF excitation pulses at the same location. As a result, signal from spins which experience either 90 or 180° (not both) due to flow will not be possible. The loss of signal is presented as "signal void."

Fig. 1B: Image obtained using a conventional spin-echo technique that uses 90° and 180° degree RF pulses in one location. Notice lack of signal in carotid and jugular vessels with flow going in both directions in this T1-weighted axial MR image as dark (The left jugular is small).
the potential in visualizing blood vessels.

Among the techniques available today, GRE imaging techniques are commonly used in which three main methods can be defined. One method uses the magnitude of available longitudinal magnetization, which is referred to as the Time-Of-Flight (TOF) technique. In the case of time-of-flight techniques, the flowing blood has higher magnetization contributing to overall signal (Fig. 2). The TOF techniques are routinely used in imaging blood vessels. The pulse repetition time, TR and flip angle are varied to optimize available magnetization for higher signal (Fig. 3). The second method uses the phase relationship among spins to highlight signal and is referred to as the phase Contrast (PC) technique. The third method, which has become very popular, is a gadolinium-enhanced MRI, which overcomes many of the difficulties of TOF and PC techniques.

All of these techniques can be performed either in a two-dimensional (2D) or a three-dimensional (3D) mode.

When flow is pulsatile, the conventional TOF or PC methods as they use only first-order cause severe periodic ghosting artifacts along low compensation scheme. Pulsatility in flow can be phase encode direction (Fig. 4). Use of ECG triggering to synchronize data acquisition with the cardiac cycle helps minimize these ghosting artifacts (Fig. 5). Use of saturation band placed superior or inferior to acquisition plane allows for selective imaging of arteries or veins (Fig. 6). We will discuss the widely used contrast-enhanced MRI and its clinical applications.

**Gadolinium contrast-enhanced 3D MRI:**

Contrast-enhanced MRI is rapidly growing in its usage. It overcomes many of the difficulties encountered with traditional non-contrast-enhanced techniques. The method relies on shortening the T1 of blood spins so that they recover nearly to their longitudinal maximum before the application of the next RF pulse.

The loss in recovery between longitudinal magnetization (due to shortened TR) can be compensated by speeding the recovery of magnetization by way of intravascular injection of paramagnetic contrast agents such as gadolinium chelates. While, the reduction in TR can be achieved either by reducing the number of phase encode steps or by increasing the bandwidth of the receiver, the
**Magnetic Resonance Angiography**

**Fig. 3:** The effect of TR on signal shown at various flip angles. As can be seen from this simulation, as TR is increased, the optimum flip angle (Ernst angle) for maximum signal shifts to a higher value. The above simulation is for normal non-contrast enhanced blood with $T1 = 1200$ ms.

**Fig. 4:** Transverse gradient recalled imaging of the neck. This pulse sequence does not have flow compensating gradients. The periodic nature of arterial pulsatile flow causes periodic ghosting artifact along the phase encode direction. The bright and dark appearance of these ghost signals is due to periodicity results in its minimum giving rise to dark appearance while the point that is at its maximum gives rise to bright signal.
Fig. 5: Imaging of the femoral arteries using a ECG triggered pulse sequence. Thirteen phase encode lines are obtained during the systolic phase of the cardiac cycle. A single stack of thirty-two overlapping 2D slices are obtained in 3 minutes. A total of 8 stacks are required to complete large field-of-view covering the femoral vessels. The total scan time is 15 minutes.

Fig. 6 (A-C), A: The effect of a saturation band to suppress signal from undesired direction of flow. (A) When no saturation band is placed, both the arteries and veins are brightly visible due to their in–flow enhancement. (B) When a saturation band is placed superior to the imaging slice, the venous signal is saturated and only arterial signal is seen. (C) When a saturation band is placed inferior to the imaging slice, signal from arteries are saturated and veins are bright.
concomitant loss in signal/noise may be compensated by the rapid recovery of blood magnetization from paramagnetic contrast. Since the signal is dependent on the recovery of longitudinal magnetization from gadolinium injection, it is independent of inflow of unsaturated spins. Thus, it is independent of plane of orientation, which is an advantage over 2D and 3D-TOF. Also with the ultra short imaging time, imaging can be performed during the breath hold and generate free from, motion related artifacts.

The pulse sequence used in CE-MRI is based on 3D GRE techniques in which TR and TE are held very short. The overall degree of T1 weighting is governed by the RF flip angle, the repetition time, TR, and the amount as well as the rate of contrast injection. The optimum flip angle may be estimated based on the desired short TR and injection rate. Most CE-MRI studies prefer short imaging during the first-pass of gadolinium contrast, as 50% of injected contrast is washed-out following the first pass. Therefore, the knowledge of the patient circulation time and bolus characteristics is an important first step to accurately image the area of interest. The key to performing rapid vascular imaging is to use a contrast injection as a bolus. The peak enhancement is visualized with the arrival of the bolus contrast during the first-pass and the peak signal is found when the bolus arrival coincides with the time corresponding to the central k-space of data acquisition. Thus, the knowledge of the bolus arrival time at the intended target is an important first step in the optimization of the protocol. This is particularly challenging in contrast-enhanced MRI, as the bolus arrival time varies from patient-to-patient depending on circulation time and cardiac output. Our approach was to use a small amount (~2ml) of contrast and use the remaining amount of the full bolus injection. A signal imaging slice is placed transversely at the region of interest and sequence parameters are used that will allow measurement of one slice per second.

With the number of repeats of the same slice set at 60, the 2 ml contrast is injected at the rate intended for the final bolus followed by a saline flush. The data acquisition is started simultaneously with the onset of injection of the small 2 ml bolus. The signal intensity at the region of interest is plotted against slice number as shown in Fig.7. The image numbers corresponding to the peak represent the elapsed time from the injection. It is important to ensure that GD concentration does not change too rapidly during its flow within a vessel. It is possible that with laminar flow and incorrect sequence timing, one will be notice ringing artifact showing dissimilar signal in the center and the boundary of the vessels. It is for this reason that the bolus duration is incorporated in estimating bolus arrival time.

More recently, the use of a stepping table has been shown for imaging of multiple body areas using a single injection. The table can be programmed to move a finite distance to relocate to a different anatomic area within the magnet center. The pulse sequence is programmed to perform three, four or five measurements, with a single measurement at each step, while the table can be moved either automatically or manually between measurements.

An automatic table stepping interface is provided by each vendor and is coupled with sequence measurement. For example, for MR angiography of the abdominal aorta, pelvic and lower extremity, first, a series of axial/ sagittal localizers are obtained at various anatomic areas. Then, at each anatomic area, a single slab is oriented in oblique coronal orientation to cover the vessels with a large field-of-view along the superior-inferior direction. Upon moving to a subsequent area, a separate 3D slab is prescribed with its own parameters.

These sets of parameters from multiple measurement slabs are stored and executed when the table starts from its first abdominal location. During the table movement, the appropriate sequence parameters are loaded and are executed when the table stops. It takes about 3-4 seconds to move the table from one area to the next.

In situations where the automatic table stepping software is not available, a manual retrofit table is used which is manually stepped from one station to the next by an operator inside the magnet room. The same test-bolus procedure is used to estimate the bolus arrival time in the abdomen station. In order to keep arterial signal high throughout the imaging time, a bi-phasic injection is used in which 1/3 of contrast is injected as a rapid bolus followed by a slow infusion of the remaining contrast which is then followed by saline flush. This allows for maintaining a higher arterial signal plateau so that
Fig. 7: Bolus Arrival Time - The images are obtained at one image per second. On the right hand side, the signal intensity of the aorta is plotted against image number. The image number corresponding to the peak signal represents the arrival time for the bolus. In this study, the injection and scan begin at the same time.

Fig. 8: Imaging of bilateral carotid arteries using a neck coil. The imaging field-of-view is 300 mm in the coronal orientation. The slab thickness and matrix is described in the text. The scan time is 11 seconds. There is an aneurysm in the right common carotid artery just before the bifurcation. A superior saturation band has eliminated any venous signal.

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signal difference between arteries and veins is increased at the distal lower extremity segments.

Clinical applications of MR angiography:

Carotid / arch vessels:

There are several natural history and epidemiological studies that demonstrate a high atherosclerotic risk factor burden in patient with known atherosclerotic disease. These include carotid stenosis disease since the prevalence of severe (greater than 60 % diameter) carotid stenosis is as high as 34% in patients with coronary artery disease undergoing Coronary Artery Bypass Graft (CABG) surgery. Diagnosis and subsequent treatment, therefore, should reduce morbidity and mortality of cerebrovascular complication of a stroke. The conventional DSA technique is implemented as a projection technique and the measurement of percent stenosis may vary by projection. With respect to patients with either peripheral vascular disease abdominal aortic aneurysms, the prevalence of severe carotid artery stenosis is 19 to 21% and increases from 7% in patients with mild renal artery stenosis to 28% with severe renal artery stenosis. Screening for carotid artery stenosis, therefore, also appears justified in this older group of patients with either peripheral and/or renal arterial disease.

Two approaches useful in the evaluation of carotid artery stenosis are the non-contrast-enhanced TOF (2D/3D) and the contrast-enhanced 3D-MRA. Using conventional 2D-TOF approaches, a series of imaging slices are perpendicular to carotid arteries and images are acquired in a sequential manner. In order to saturate spins from jugular veins, a traveling saturation slab is placed superior to acquisition slice. The total acquisition time can be as long as 15 minutes. Typically, the parameters used in a 2D-TOF are as following:

TR (msec) / TE (msec) / Flip angle (deg) = 608/7/70
Field-of-view = 175 (phase) x 350 (frequency)
Matrix size = 114 x 256
Resolution (mm$^2$) = 1.54 x 1.37x3
Scan time = 2 min 57 sec,

The second approach requires a test bolus procedure as described in text. A 3D-MRA pulse sequence is setup in a coronal orientation and is initiated at the estimated delay from the injection. Multiple measurements are performed to detect arterial anatomy as well as venous anatomy using these parameters:

TR (msec) / TE (msec) / Flip angle (deg) = 3.37 / 1.22/25,
Field of view = 341 (phase) x 390 (frequency)
Matrix size = 246 x 512,
Resolution (mm$^2$) = 1.4 x 0.8 x 1.2,
Scan time = 20 sec.

Contrast amount = single dose (0.1 mmol/kg) and injection rate = 2.5 ml/sec.

Figure 8 shows an MIP of a single data set acquired in coronal orientation. Figure 9(A,B) shows thoracic and arch vessels obtained using a back-to-back data arterial phase and the second shows venous signal.

Thoracic aorta:

Among the number of common causes of thoracic aortic disease, atherosclerosis, congenital anomalies and inflammatory processes are prominent.

Some of them are life threatening and may need immediate attention such as dissection. Although, CT has wide spread use in evaluating the thoracic aorta for acute aortic trauma, aneurysms and aortic dissections, MRA and MRI are reserved for problem solving and as the modality of choice for congenital abnormalities. Also, patients with renal insufficiency are imaged with MRA/MRI instead of with CT. For a comprehensive MR evaluation of the thoracic aorta, 3D-MRA is also included in the imaging protocol. The technique is based on the same physical principle described earlier, which is independent of flow direction. Also, it is not affected by slow flow and, therefore, can easily be used to differentiate thrombus from slow flow. The pulse sequences are rapid and a volume of image data can be obtained in under 10 seconds with stronger gradients. Due to the rapid nature of the imaging pulse sequences, for the most part, ECG triggering is not required. Flow dynamics may be evaluated by supplementing CE-MRA with a 2D cine study using phase-contrast techniques. However, this would require the placement of ECG leads to synchronize cine phase-contrast data.
acquisition with the cardiac cycle. This technique, however, can supplement the CE-MRA study to depict the morphology and function of the aortic valve. For thoracic MRA, patients lie in the supine with a dedicated phased-array coil covering the chest. A posterior spin coil elements cover the patient’s thoracic aorta. Before patient centering, an IV line is started at the anti-cubital fossa. A test-bolus is performed to estimate the bolus arrival time. Imaging is performed in a sagittal or an oblique sagittal orientation with multiple measurements in a single breath hold.

The pulse sequence parameters are:

- TR (msec) / TE (msec) / Flip angle (deg) = 3.37 / 1.22 / 25.
- Field-of-view = 341 (phase) x 390 (frequency).
- Matrix size = 246 x 512.
- Resolution (mm) = 1.4 x 0.8 x 2.0
- Scan time = 14 sec.
- Contrast agent = 0.1 mmol/kg at an injection rate of 2-3 ml/sec followed by saline flush.

Figure 10 shows an MIP from data obtained in a sagittal orientation in a patient with coarctation.

**Abdominal Aorta:**
The primary use of 3D contrast-enhanced MRI in the abdomen has been to evaluate the abdominal aorta and its visceral branches, for aortic aneurysms and visceral branch stenosis. Patients are positioned in a supine orientation with an IV line at the antecubital fossa. A phase array coil is ideally suited for signal reception. A test bolus injection is used to estimate the bolus arrival time. Normally, a single-dose (0.1 mmol/kg) is used with a 2-3 ml/sec injection rate. The diagnostic analysis must also accompany source images and multi-planar reformatted images. Figure 11 shows a sagittal MIP projection of data obtained using a sagittal 3D slab. Occlusion of the celiac artery and distal aorta with SMA stenosis is clearly visible.

**Renal arteries:**
The pulse sequence for evaluating the renal arteries is similar to the other techniques used in CE-MRA studies. 3D-MRA may be complemented with a flow quantification study in order to assess the degree of stenosis. This is especially helpful in evaluating for stent where the signal loss due to the stent does not
reveal vascular pathology in a contrast-enhanced study.

The patient is positioned inside the magnet with a phased array coil at the kidney area. The IV line is started at the antecubital fossa. The test-bolus study is performed at the level of the renal hilum. With proper timing, multiple 3D-MRA measurements are performed in a single breath-hold period. The following pulse sequence parameters are used:

TR (msec) / TE (msec) / Flip angle (deg) = 3.37 / 1.22 / 25
Field-of-view = 341 (phase) x 390 (frequency),
Matrix size = 246 x 512,
Resolution (mm²) = 1.4 x 0.8 x 2.0,
Scan time = 14 sec,
Contrast agent = 0.1 mmol/kg with an injection rate of 2-3 ml/sec followed by saline flush.

Figure 12 shows an MIP image of subtracted data set (post-mask). The resulting images shows the renal arteries and a severe stenosis of the left renal artery.

**Mesenteric circulation:**
Contrast-enhanced 3D MRI is highly successful in evaluating patients with mesenteric ischemia. The spatial resolution is sufficient to accurately evaluate the origins of splanchnic arteries, superior mesenteric artery and celiac axis. Patients presenting with mesenteric ischemia are usually thin and can easily be imaged with a body phased array coil for better signal reception. In this study, either a straight sagittal or an oblique coronal plane is performed following contrast injection. The origin of SMA is seen clearly on a sagittal presentation (Fig. 11) while the distal branches are seen on an oblique coronal presentation (Fig. 12). Figure 12 shows an MIP of data obtained using an oblique coronal slab position for data acquisition with the following parameters:

TR (msec) / TE (msec) / Flip angle (deg) = 3.37 / 1.22 / 25.
Field-of-view = 341 (phase) x 390 (frequency),
Matrix size = 246 x 512,
Resolution (mm²) = 1.4 x 0.8 x 2.0,
Scan time = 14 sec,
Contrast agent = 0.1 mmol/kg with an injection rate of 2-3 ml/sec followed by saline flush.

**Peripheral vascular imaging:**
Currently, the most common application for stepping table MR angiography is for the peripheral lower extremities to assess renal artery stenosis, peripheral vascular disease and/or graft patency.

The combined use of multi-station MR angiography and MR imaging is also very useful in patients with graft infections. In the future, lower extremity MR venography may also supplement peripheral lower extremity MR angiography for mapping of the saphenous veins and may also supplement pulmonary MR angiography in patients suspected of having thromboembolic disease. Also, whole body cardiovascular screening appears to be, not only feasible, but a very promising new application of stepping MR angiography tables.

Imaging of the entire lower extremity begins from abdomen or skull-base is possible with a moving table. In this technique, the patient anatomy is repositioned to the subsequent station following data acquisition at that station. The technique allows one to chase the bolus from the abdomen or chest as it travels down to lower extremity. Rapid repositioning is the key to performing multistation MRA using a single injection.

The patient is placed on a moving table in the supine position, which is retrofitted on the existing MR patient able. This sliding table positioned with the patient sandwiched between an anterior and a posterior phased array coil, which is held fixed with respect to the magnet center. The different anatomic areas will be brought to the imaging coil and magnet center as the patient is moved outward. Patient will be started with an IV line at the antecubital fossa. The imaging protocol involves a test bolus at the proximal station. Imaging is performed in a coronal orientation using the following parameters:

TR (msec) / TE (msec) / Flip angle (deg) = 4.6 / 1.8 / 25
Field-of-view = 341 (phase) x 390 (frequency)
Matrix size = 246 x 512
artefactual vasculature. This imaging
Resolution = 1.4 x 0.8 x 2.0,
Scan time = 18 sec
A single measurement as achieved in 18 seconds
With the number of measurements set equal to 3 and
with a 3 second pause between measurements, the
Fig. 10: Thoracic aorta is imaged using a phased array artery coil. The imaging parameters are described in the text. As can be seen there is a curtailation of the aorta resulting in prominent intercostals and internal mammary arteries.

Fig. 11: Abdominal aorta imaged using a phased array body coil. The imaging parameters are described in the text. In this sagittal plane, MIP shows marked narrowing of the proximal superior mesenteric artery. Also, there is almost complete occlusion of distal abdominal aorta and proximal celiac artery.
Fig. 12: Abdominal aorta including renal arteries. This imaging is performed using a phased array coil. The imaging parameters are described in the text. Severe stenosis of the left renal artery is seen. Note the advanced degree of atherosclerosis of the abdominal aorta.

Fig. 13: Splanchnic arterial vasculature. This image is performed using a phased array coil. The imaging parameters are described in the text.
Magnetic Resonance Angiography

Fig. 14: Lower extremity run-off vessel imaging with a SKIP stepping table. The table increment is fixed at 40 cm and the image FOV is kept at 50 cm allowing for 5 cm overlap. Patient with a severe occlusive disease of bilateral femoral and anterior tibial arteries.

Fig. 15: Whole-body MRA beginning from skull-base to feet using SKIP stepping table. A total of 5 stations cover from 130 cm with a table increment step of 30 cm. The measurement FOV was 40 cm with an overlap of 10 cm.

First measurement typically begins at the level of abdomen at the desired delay time from injection. A biphasic injection is performed in which 15 ml is injected @ 1.5 ml/sec followed by remaining contrast (based on 0.2 mmol/kg) at 0.9 ml/sec, which is followed by saline flush of 20 ml at 1.5 ml/sec. Figure 14 illustrate the application of this method. Severe occlusive disease of bilateral femoral and tibial arteries are seen. Finally, Fig 15 demonstrate application of table stepping using SKIP (25) for whole body cardiovascular screening.

Conclusion:
Magnetic Resonance Angiography has evolved into the modality of choice for screening or initial diagnosis and evaluation of many clinical conditions. Patients with suspected cerebrovascular, thoracic or abdominal aortic disease such as atherosclerosis, aneurysm, dissection or congenital abnormalities can benefit from this non-invasive approach. Also, patients with intestinal angina or hypertension when the respective underlying cause may be compromised mesenteric circulation or stenosis of renal arteries can easily be assessed by MRI. While the technology for performing MRI continues to improve, there is parallel expansion in the clinical applications of this invasive, easily to perform highly rewarding diagnostic study.

References:


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