Chapter 1

Flow

1.1. Blood

1.1.1. Characteristics of bloodflow

Bloodflow is characterized by four parameters:
– its velocity;
– its acceleration;
– its direction;
– its orientation.

Arterial flow has the highest velocity.

This velocity varies with the phases of the cardiac cycle: it may be very high (150-175 cm/s during systole, the period of contraction of the myocardium) but also almost null (at the end of diastole, the period of relaxation of the myocardium).

The flow in the cerebral arteries is much slower: 40-70 cm/s.

Finally, venous flow is the slowest: generally less than 20 cm/s.

Bloodflow runs from the organs to the heart in venous circulation and from the heart to the organs in arterial circulation.
1.1.2. Laminar flow and turbulent flow

We can distinguish two main types of flow:

– A laminar flow is found when the velocities are relatively low. It is characterized by a distribution of velocities which are all parallel and whose profile is parabolic: the velocity of the fluid is maximum at the center and almost null when against the walls.

![Figure 1.1. Profile of velocities in a laminar flow](image)

– A turbulent flow is a chaotic form of flow. The velocity then exhibits a vortical nature: local circular motions arise.

![Figure 1.2. Profile of velocities in a turbulent flow](image)

There are many reasons why a laminar flow could become a turbulent flow:

– the velocity of the flow surpasses a critical value;
– the structure of the vessels changes (see Figure 1.2).

Venous flow is laminar with a velocity that is constant overall.

Arterial flow is an intermediary case: it is laminar in diastole and turbulent during systole.

A turbulent flow causes significant artifacts in images of the flow, which we shall now go on to discuss in detail.
1.2. Basic phenomena in angiography

The flow phenomena which we are going to describe in this section enable us to spontaneously view blood vessels (without the injection of a contrast-enhancing product); the effect they have on the image depends on the sequence used (spin echo (SE) or gradient echo (GE)), on the parameters of that sequence (T\textsubscript{R}, T\textsubscript{E}...), but also on the particular parameters of the flow itself: its velocity, the orientation of the vessel in relation to the slice, etc.

Thus, we can construct a form of imaging relating to the flow: magnetic resonance angiography, or MRA.

1.2.1. Time of Flight (TOF)

When the vessel runs through the slice, the intensity of the flow signal depends on the time-of-flight of the protons, i.e. the time taken to traverse the thickness Δz of the plane being imaged at velocity V.

1.2.1.1. Phenomenon of flow void in a spin-echo sequence

We work on a spin-echo sequence where the 90° and 180° pulses are selective in the given slice.

*With stationary protons*: they experience both pulses and are therefore able to generate a signal.

*With moving protons* (in the bloodstream), two scenarios may arise:

- Either they remain in the slice (T\textsubscript{t} < T\textsubscript{E}/2) and are therefore struck by both pulses to generate a signal. For a vessel perpendicular to the slice, these protons have a velocity of V < Δz/(T\textsubscript{E}/2): this is qualified as a “slow” flow.

- Or they leave the slice entirely before the transmission of the 180° pulse (i.e. T\textsubscript{t} < T\textsubscript{E}/2). They are then replaced within the slice by protons which were not subjected to the initial 90° pulse and which therefore do not generate a signal. (The extreme case, T\textsubscript{t} = T\textsubscript{E}/2, is represented in Figure 1.3). For a vessel perpendicular to the slice, these protons have a velocity of V > Δz/(T\textsubscript{E}/2): this is qualified as a “fast” flow.

*NOTE*: “Fast flow” is not necessarily synonymous with “arterial flow”: flow in large veins is also included in this category.
Flow void is the most commonly encountered phenomenon in flow imaging: this phenomenon accounts for the fact that many blood vessels are spontaneously visible on an MRI image by the absence of signal (by the “wall / light” contrast).

1.2.1.2. Phenomenon of flow-related (or paradoxical) enhancement

1.2.1.2.1 The phenomenon

We work on a spin-echo or gradient-echo sequence.

We suppose the $T_R$ to be short: for this reason, there is repetition within a sufficiently short time period of the 90° excitation pulses.

*With stationary protons:* they have already been excited by the 90° pulse from the previous cycle (i.e. from the previous $T_R$ interval): therefore, their longitudinal magnetization is low and they generate little signal.

*With moving protons:* if their velocity is sufficient, “fresh” protons, which have not been subjected to the 90° pulse from the previous cycle, enter into the slice: their longitudinal magnetization vector is maximum and the signal engendered by these protons is therefore maximum as well: this is the phenomenon of paradoxical enhancement.

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**Figure 1.3.** Case of $T_i = T_E/2$: the fate of protons experiencing the 90° excitation pulse (in white dots) and cross-sectional image of the flow
1.2.1.2.2. The $T_R$ dictates this phenomenon

The phenomenon of paradoxical enhancement is all the more marked when the TR is short: gradient-echo sequences (see below) are therefore more sensitive to this phenomenon, and therefore we shall limit ourselves to this type of sequence in our discussion below.

**Figure 1.4.** Case of $T_i > T_R$: fate of protons in the flow that experience the $90^\circ$ excitation pulse (in white dots) and cross-sectional image of the flow. The vertical slab indicates all of the protons in the flow (in black dots) or outside of it (in white) affected by the different pulses.

**Figure 1.5.** Case of $T_i > T_R$: longitudinal and transversal magnetization of the protons in the flow (top) and outside of it (bottom).

This phenomenon is maximum when all the protons excited during a cycle leave the slice and are replaced by “fresh” protons in the next cycle ($T_i < T_R$).
For a vessel running perpendicular to the slice, these protons have a velocity of $V > \Delta z/T_R$. (The extreme case, with maximum contrast: $T_i = T_R$, is represented in Figure 1.6).

\[ RF \quad \text{T}_R \quad 90^\circ \quad \text{T}_R \quad 90^\circ \]

**Figure 1.6.** Case of $T_i = T_R$: fate of the protons in the flow that experience the 90° excitation pulse (in white dots) and cross-sectional image of the flow. The vertical strip indicates all the protons in the flow (in black dots) or outside of it (in white) affected by the different pulses.

**Figure 1.7.** Case of $T_i = T_R$: longitudinal and transversal magnetization of the protons in the flow (top) and outside of it (bottom).

**Note.** For spin-echo sequences, if $T_R = 1s$ and $\Delta z = 1$ cm, then paradoxical enhancement is maximum for velocities $V > \Delta z/T_R$, i.e. $V > 1$ cm/s (this velocity, which is very slow, corresponds to venous flow at the end of diastole). The contrast
obtained (the difference between flows of different velocities) is therefore very poor – all the more so if the stationary protons in the tissue are not saturated with a similarly long $T_R$.

This indeed reaffirms the necessity of working in a gradient-echo regime if we wish to use the paradoxical signal.

1.2.1.2.3. Single slice and multi-slice

In a multi-slice sequence, the paradoxical signal predominates over the first slice(s) (or the last) depending on the orientation of the flow (i.e. on the “flow entry slices”): such is the case for slice A in the diagram below.

Indeed, in planes which are a long way from the flow entry slices, the protons have been saturated by previous $90^\circ$ excitations in the upstream slices (much like stationary protons): such is the case for slices C, D and E.

However, the paradoxical signal may also be visible on the first intermediary slices (slice B): sometimes we note the remnant presence of an intravascular “halo” signal in these slices. This halo is due to the differences in velocity of the protons at the center of the vessel (fast flow) and at the edges of the vessel (slow flow) in the case of a laminar flow with parabolic profile.

![Figure 1.8. Paradoxical enhancement in a multi-slice sequence](image)

NOTE.— The gradient-echo sequence, which is favored when we wish to use the paradoxical signal, usually enables us to take very short $T_R$ values, which generally lends itself to single-slice processing: thus, each successive slice behaves like a new “entry slice”.
1.2.2. Phenomenon of dephasing of circular spins

1.2.2.1. Dephasing of protons in a unipolar gradient

We consider protons whose velocity $v$ is supposed to be constant (i.e. with zero acceleration) and direction $\mathbf{x}$. The phase of the protons will depend only on the gradient in that direction: $G_x$.

At a time $t_e$ taken as a reference point, the position of the protons is $x(t_e)$ and their velocity in the direction of $G_x$ is $v(t_e)$.

We wish to discover the phase $\phi(t_m)$ of the protons at a time $t_m > t_e$.

Thus, let us take a gradient of time-length $\tau$ and constant amplitude $A$.

The gradient is switched on at time $T - \frac{\tau}{2}$ and switched off at $T + \frac{\tau}{2}$.

Thus $\phi(t_m) = \int_{T - \frac{\tau}{2}}^{T + \frac{\tau}{2}} \gamma \cdot G_x \cdot x(t) dt = \int_{T - \frac{\tau}{2}}^{T + \frac{\tau}{2}} \gamma \cdot G_x \cdot [x(t_e) + v \cdot (t - t_e)] dt$.

Therefore: $\phi(t_m) = \gamma \cdot A \cdot \left( x(t_e) \cdot \tau + \left[ \frac{v \cdot (t - t_e)^2}{2} \right]_{T - \frac{\tau}{2}}^{T + \frac{\tau}{2}} \right) = \gamma \cdot A \cdot (x(t_e) \cdot \tau + v \cdot (T - t_e) \cdot \tau)$

However, $x(T) = x(t_e) + v \cdot (T - t_e)$, so $\phi(t_m) = \gamma \cdot A \cdot x(T)$

Note.— The phase is independent of $t_m$.

1.2.2.2. Dephasing of the protons in a bipolar gradient

Based on this simple gradient, we can construct gradients which behave differently in relation to stationary or moving protons.
The amplitudes of the two lobes of the bipolar gradient shown in Figure 1.9 are respectively $A$ and $-A$.

The phase obtained at time $t_m$ is thus given by:

$$\varphi(t_m) = \gamma \cdot A \cdot x(T_1) - \gamma \cdot A \cdot x(T_2) = -\gamma \cdot A \cdot (T_2 - T_1)$$

Hence:

$$\varphi(t_m) = -\gamma \cdot v \cdot A \cdot \tau'$$

For stationary protons: $\varphi = 0$, the protons are refocused.

(NOTE.– we quite deliberately discount the dephasing due to the inhomogeneities of the field $\varphi_B$ and to the Foucault currents $\varphi_e$.)

For moving protons, their movement in the direction of the gradient makes it impossible to re-phase their dephasing: their phase is proportional to their velocity.
NOTE.– it is not necessary for the two lobes of the gradient to be adjacent: the time $\tau'$ is, in this case, the length of time between the centers of gravity of each of the two lobes, and the formula $\phi(t_m) = -\gamma \cdot \mathbf{V} \cdot A \cdot \tau'$ remains valid.

1.2.2.3. Form of velocity-insensitive gradients: flow compensation gradient

The frequency gradients (selection of slice and readout) mark protons when they are applied. Generally, they are bipolar in form, to facilitate the protons' readjustment phase.

However, this rephasing is possible only for stationary protons, rather than moving protons, whose dephasing is due to their velocity, as shown by section 1.2.2.2.

In order to compensate for the dephasing due to the displacement of protons circulating at constant velocity, we need to add lobes to the frequency gradients.

In order to do this, the bipolar gradient is altered to possess three lobes in the ratio 1:2:1.

![Figure 1.11](attachment:image.png)

**Figure 1.11. Alteration of the bipolar gradient to compensate for the phase shifting due to the displacement of protons and thereby obtain a flow compensation gradient**

This type of gradient is indeed the sum of two bipolar gradients; the first creating a phase-shift $\phi_1(t_m) = -\gamma \cdot \mathbf{V} \cdot A \cdot \tau'$ of the moving protons, and the second an opposite phase-shift: $\phi_2(t_m) = +\gamma \cdot \mathbf{V} \cdot A \cdot \tau'$. Finally, $\phi_{TOTAL}(t_m) = 0$: there is rephasing of the stationary protons and the moving protons.
Flow

Figure 1.12. Phases acquired by the different protons when a flow compensation gradient is applied

**NOTE.**—Obviously, it is also possible to compensate for the dephasing caused by constant acceleration. We need only add extra lobes.

However, the disadvantage of adding lobes to the reading gradient is that it prolongs the $T_E$, which also entails an increase in the dephasing of the spins (a decrease in $T_2^*$) and therefore weakens the signal.

For this reason, only phase-shifts linked to a constant velocity tend to be corrected, with $T_E$ being kept as short as possible.

### 1.3. Artifacts relating to the flow

The artifacts described in this section are those which can be “conventionally” encountered, without necessarily intending to perform angiography. (Thus, the artifacts relating to MRA sequences, such as aliasing, are discussed later on in the section relating to the sequence itself).

Here, therefore, we are interested in “conventional” artifacts of flow.

#### 1.3.1. Artifact of pulsatile flow of blood or cerebrospinal fluid (CSF)

##### 1.3.1.1. Causes...

The varying velocity of CSF or blood over the course of the cardiac cycle means that the intensity of the signal measured at the same voxel will be periodic over time:
– the renewal of non-saturated protons is surely to be found during the period of systole (high average flow velocity) and absent during diastole (low flow velocity);
– the dephasing of spins in the same voxel may be great during systole (high average flow velocity, so high velocity contrast and therefore phase contrast) and low during diastole (low flow velocity, so low velocity contrast and therefore phase contrast).

1.3.1.2. …and consequences

Thus, the pulsatile flow of blood or CSF causes artifacts which often take the form of ghost images in the direction of phase coding, the intensity of which is variable depending on the extent of the effects of TOF or dephasing of the moving protons.

It is possible to see these artifacts in the case of a spin-echo sequence: during systole, the moving protons excited by the 90° pulse exit the slice and are therefore not touched by the 180° pulse; during diastole, however, the signal is strong because the average velocity of the protons is low.

Nevertheless, these artifacts are often more pronounced on gradient-echo sequences, where because of paradoxical enhancement, the intensity of the signal may be very great.

We regularly find this type of artifact in transverse slices containing the aorta, as shown in Figure 1.13.

![Figure 1.13](image)

**Figure 1.13.** Axial view of the abdomen. Flow artifact in the direction of the phase coding gradient due to the paradoxical enhancement of the signal from the vena cava (1) and the aorta (2)

It is important to reduce the impact of this artifact, as a diagnosis (such as the detection of a vertebral angioma, for instance) could be made difficult if ghost images are projected onto the image of interest.
1.3.1.3. Solutions

There are various ways to decrease ghost images:

– *ECG synchronization* can be employed to always trigger the acquisitions at the same moment in the cardiac cycle, which reduces signal modulation due to a difference in the average velocity of the bloodflow in the vessels.

– *Flow compensation gradients* minimize pulsatile artifacts in the blood or CSF by reducing the dephasing due to the fairly rapid displacements of the protons within the same voxel.

– *Pre-saturation* involves applying saturation slabs to the slices upstream of the one under examination.

![Figure 1.14.](image)

*Figure 1.14. Pre-saturation strip (white) and its effect on the flow signal*

The role of these slabs is to saturate the blood in the vessels before it arrives at the slice of interest, so that the moving protons do not emit a signal (hence there can be no modulation of the intensity of that signal).

![Figure 1.15.](image)

*Figure 1.15. Elimination of the artifact of flow encountered previously (see Figure 1.14), using pre-saturation slabs applied... above the slice: elimination of the ghost image of the aorta but not of the vena cava (top left) ... below the slice: elimination of the ghost image of the vena cava but not of the aorta (top right)... above and below the slice: elimination of both artifacts (bottom)*

NOTE.– Pre-saturation also enables us to select a flow on the basis of its orientation: for instance, we could eliminate venous flow from consideration, keeping only arterial flow.

– Finally, we can *invert the gradients of phase coding and frequency*: this does not prevent ghosting from happening, but moves the ghost image to a different area in the image of interest. Thereby, we are able to be sure that it is indeed an artifact rather than a diseased structure.
1.3.2. Fluid location error

1.3.2.1. Description

In an SE or GE sequence, the encoding in the direction of the phase gradient is done before that of the frequency gradient. This causes an error in locating the moving protons, as we shall shortly see:

- Phase coding takes place after a time $T$ after the excitation pulse; we shall consider the moment when the row $k_y = n$ is acquired. The increment of the gradient $G_y$ and its duration are respectively notated as $\Delta G_y$ and $\tau$. The phase of the protons which at that time are in position $(x(T), y(T))$ is therefore:

$$\varphi = \gamma \cdot n \cdot \Delta G_y \cdot y(T) \cdot \tau$$

- At the time of the gradient echo $G_x$, i.e. at time $T_e$, the protons are displaced and are now located in position $[x(T_e), y(T_e)]$, but are recorded in position

$$[x(T_e), y(T_e)] = [x(T_e), y(T_e) - v_y \cdot (T_e - T)]$$

[1.1]

The signal is therefore shifted in the direction of the phase gradient: we then see the appearance of a row with no signal, which is the vessel, bounded by a row with intense signal: that of the protons of the blood.
1.3.2.2. Parameters influencing the artifact of location

1.3.2.2.1. Orientation of the flow

For most SE or FE sequences used, it is possible to determine the orientation of blood circulation when an artifact of location is shown.

Indeed, we can see that the signal row is positioned alongside the signal-free row given by the component of the flow in the direction of the frequency gradient (indicated by the dotted arrow in the diagrams below).

This artifact of location is therefore widely used in clinical practice, particularly in cases of vascular malformation.

1.3.2.2.2. Echo time

The error in location of the flow is greater for vessels situated within the slice being examined. It is accentuated when the echo time $T_E$ is increased, as we can see from equation [1.1].

![Figure 1.17. Position of the artifact of location and orientation of flow](image)

1.3.2.3. Clinical reality

As is indicated by equation [1.1], the blood signal shifts proportionally to the component of velocity of the flow along the phase gradient axis: $v_y$. 
As vessels are not linear, this component varies with the position of the fluid: the artifact of shift is therefore also dependent on the position of the fluid (for instance, it is non-existent if $v_y = 0$); hence, there is more than a simple displacement: there is distortion of the blood vessels subjected to this artifact.

![Figure 1.18. Distortion in the case of a gradient-echo sequence showing the anterior cerebral artery. These images are obtained with identical echo times, but different directions for the phase gradients (left to right (LR) in the first image above and anterior to posterior (AP, front to back) in the second). (From [VLA03]. Reproduced with kind permission from Springer)](image)

If we compare the images in Figure 1.18, obtained with a gradient-echo sequence, we note that the incurvate form of the anterior cerebral artery varies.

In the upper image, the distance between the two branches of the incurvate part is greater: this results from a shift of these two branches in the direction of the phase gradient (LR direction) and in opposite orientations.
In the image below, the displacements of these two branches are in the AP direction: the two branches are shifted downwards (this shift is greater when the component $V_y$ is large, i.e. the bifurcation peak; this gives the peak a squashed appearance).

**Figure 1.19.** Distortion in the case of an LR phase gradient. The orientation of the flow is shown by the velocity vectors in solid line. The dotted vectors indicate the displacement of the signal.

**Figure 1.20.** Distortion in the case of an AR phase gradient. Notations identical to those in Figure 1.19.
1.3.2.4. Solution

To correct the artifact of location is to correct the phase given by the application of the gradient $G_y$, as though the gradient had in reality been measured at time $T_E$.

- Without correction, at time $T$, the phase is: $\phi(T) = \gamma \cdot n \cdot A G_y \cdot y(T) \cdot \tau$

- At time $T_E$, the phase is:

$$\phi(T_E) = \gamma \cdot n \cdot A G_y \cdot y(T_E) \cdot \tau = \gamma \cdot n \cdot A G_y \cdot \left(y(T) + v_y \cdot (T_E - T)\right) \cdot \tau$$

Therefore,

$$\phi(T_E) = \phi(T) + \gamma \cdot n \cdot A G_y \cdot v_y \cdot (T_E - T) \cdot \tau$$

In order for it to seem to have been measured at time $T_E$, we should have $\phi(T) = \phi(T_E)$.

Thus, to $\phi(T)$ we need to add the correction $\Delta \phi = \gamma \cdot v_y \cdot n \cdot A G_y \cdot (T_E - T) \cdot \tau$.

However, we know (see section 1.2.2.2) that in the case of a bipolar gradient of duration $\tau$ and applied in direction $y$, the phase obtained is $-\gamma \cdot v_y \cdot A \cdot \tau$.

Thus, during the coding of the row $k_y = n$, we need to add to the “normal” phase gradient a bipolar gradient of duration $\tau$ and amplitude $-n \cdot A G_y \cdot (T_E - T)$.

We note that it is thus possible to correct the artifact of location for a flow with unknown velocity (as the above formula makes no reference to that velocity).

**Note.** This phase correction is costly in terms of time. Obviously, if a relatively short echo time is chosen, this correction cannot be done.

1.3.3. Other artifacts

1.3.3.1. Artifact due to a velocity gradient

1.3.3.1.1. Causes and consequences

On the walls of the vessels, there is a significant velocity gradient (see 1.1.2); as indicated in section 1.3.1 (ghost images due to pulsatile flow), this causes significant...
dephasing within the same voxel, and therefore a loss of signal on the periphery of the vessel. Its diameter therefore appears reduced; this artifact manifests itself mainly on axial slices.

![Figure 1.21. Vessel without (left) and with (right) a velocity gradient](image)

This intravoxel phase dispersion is also observed when there is turbulence or acceleration of the flow (e.g. in cases of stenosis) or in the vicinity of a magnetic field gradient.

1.3.3.1.2. Solution

A flow compensation gradient can considerably reduce this artifact (the same way it reduces the artifact due to pulsatile flow).

1.3.3.2. Another artifact encountered in an SE sequence

With an SE sequence with multiple (and symmetrical) echoes, a special case is represented by the rephasing of the protons on even-numbered echoes.

Indeed, the dephasing of the protons, due to the flow and observable with the first echo, is exactly compensated by the second $180^\circ$ pulse. Thus, the signal is attenuated in the first echo, but in the second echo (and all symmetrical even-numbered echoes), we shall see “recuperation” of the signal.

![Figure 1.22. Intensity of the different echoes obtained in the presence of a flow](image)
1.4. MRA sequences

The physical phenomena described in section 1.2 form the basis of the technique of MRA.

Depending on the underlying phenomenon, these techniques will be more or less sensitive to certain velocities of flow, or to certain types of flow. In particular, they will run into difficulty when faced with complex or turbulent flows.

1.4.1. Time of Flight (TOF)

1.4.1.1. Technique

1.4.1.1.1. Principle

In time-of-flight MRA, we use gradient-echo sequences to favor the signal from the flows by saturating the signal from stationary tissues with very short TR times: thereby, the longitudinal magnetization of these tissues does not have time to recover, and the signal generated by them fades, encouraging the phenomenon of slice entry: as the circulating blood entering into the slice being examined has not been saturated, its longitudinal magnetization is maximum. The signal from the blood flow is therefore stronger than that from the stationary tissues: this is known as “white blood” imaging (see section 1.2.1.2).

![Figure 1.23. Circle of Willis in white-blood TOF imaging (taken from [IRM])](image)

It is also possible to use spin-echo sequences: the circulating blood flows out of the slice; therefore it is not exposed to the selective 180° pulse and consequently is not refocused: it does not yield a signal. This is the phenomenon of flow void, which is at the root of “black blood” imaging (see section 1.2.1.1).
1.4.1.1.2. Choice of parameters

The strength of the vascular signal depends on:

– the velocity and type of flow;

– the length and orientation of the vessel being explored (the vascular signal will be stronger if the slice is perpendicular to the axis of the vessel);

– the parameters used for the sequence:

  - the repetition time TR is often chosen to be as short as possible to enhance the contrast due to paradoxical enhancement,

  - the thickness of the slice $\Delta z$ (see section 1.2.1.2.2),

  - but also the *flip angle* (the saturation of the stationary tissues is faster if the flip angle is wide).

1.4.1.1.3. A 3D representation of flow: the MIP algorithm

Maximum Intensity Projection (MIP) consists of projecting into a plane the whole of the 3D volume acquired (by a 2D or 3D sequence).

The signal is analyzed throughout the volume, and the maximum intensity pixels (above a certain predefined threshold) are projected onto a 2D image.

In order to obtain a complete volumetric reconstruction, we need to operate on slices whose projection axis is incremented (by 10° to 15°).

All of the 2D images obtained by projection can be viewed sequentially in an animated loop, giving the illusion of a true 3D image which can be rotated and viewed from any angles of incidence.

![Figure 1.24. Principle of MIP reconstruction](image_url)
1.4.1.2. Limits and optimizations

The main limitations of TOF MRA are:

– *loss of signal* when the flows are too *slow* or are oriented in parallel to the slice;

– *loss of signal* when the flows exhibit a *velocity gradient*: we then observe a reduction of the size of the vessels (see section 1.3.3.1); in order to deal with this problem we can use flow compensation gradients;

– *poor elimination of the signal from the stationary tissues with a short T₁* (primarily fat but also hematomas and thrombi), which maintain a relatively strong signal even with short TR times: they may, in certain cases, have a signal similar to that of a circulation flow lesion such as an aneurism, which skews the diagnosis.

*With regard to fat, we can then:*

– use sequences that are able to selectively eliminate the signal from the fat (such as STIR or FLAIR sequences);

– use a *magnetization-transfer* preparatory pulse: this technique is based on the fact that free protons and bound protons (belonging to macro-molecules such as fats or proteins) do not have the same resonance frequency (there is a difference of 1500 Hz). Thus, if we use a preliminary sequence that saturates the bound protons (choice of the associated frequency for the excitation pulse), those protons transfer their magnetization and thereby saturate the free protons in turn. White and gray substances, with a high concentration of bound protons, will therefore only emit a relatively weak signal. Conversely, the blood, with a low concentration of bound protons, will emit a strong signal.

*With regard to hematomas or thrombi,* there is no real “remedy”: the best we can do is to favor the use of an MRA sequence in phase contrast (see section 1.4.2):

– *artifacts in MIP and placement of the acquisition volume.*

The MIP algorithm has a *threshold* below which the signal is considered null and is therefore not projected. Thus, vessels of small diameter, whose signal is weaker than the required threshold, are not visualized by projection. In this case, analysis of the slices acquired initially (the “native slices”) is necessary so as not to miss any vessels.

Another artifact is due to the *placement of the acquisition volume*: when the volume is badly positioned in relation to the vessels needing to be explored, one or more vessels may be situated outside of the volume acquired. This gives us an image of false stenosis or false thrombosis on MIPs.
In order to detect this type of artifact, it is necessary to visualize projections obtained at different angles (e.g. a coronal and a sagittal approach).

![Figure 1.25. Defective MIP reconstruction detected by visualization of a coronal projection (left) and a sagittal projection (right): the acquisition box is located too far to the rear, and generates a false thrombosis image](image)

1.4.1.3. Different types of acquisition of a volume

When we wish to use MRA, there is an almost automatic necessity to acquire a volume (to be aware of the vascularization of an area) and therefore not to limit ourselves to a single slice.

There are two possibilities for acquiring a volume: 2D or 3D acquisition.

1.4.1.3.1. 2D TOF (Time Of Flight) sequence

In 2D acquisition, TOF imaging is performed using a series of slices acquired one after another, and stacked together to reconstruct a 3D volume.

For better contrast, the slices need to be placed perpendicularly to the main trajectory of the vessel(s).
Advantages

- Relatively fast acquisition (5 to 10 minutes, depending on the resolution and the thickness of the volume being imaged).
- Apt for different types of flows, even with slow flows (even in this case, it is possible to achieve a relatively good contrast).
- Possibility of using wide flip angles (which gives better saturation of the stationary tissues and a stronger vascular signal).

Disadvantages

- Poor spatial resolution on the axis of stacking of the slices, because the slices are fairly thick (d > 2mm);
- The poor resolution of the voxels on the axis of stacking causes a loss of signal in areas where the flow is turbulent (wide dispersion of phases).
The same problems arise when the flows are oriented in parallel to the slice: the size of the vessels is reduced, and stenoses (which are visible by the absence of signal) may therefore be overestimated.

Figure 1.28. Loss of signal in a horizontal portion of the right anterior tibial artery because of saturation of the flow parallel to the slice. (From [VOS 98]. Reproduced with kind permission from Elsevier)

Note.– this problem is worsened in MIP imaging. Indeed, a weak vascular signal (below the specific threshold of the projection) will not appear, which causes a danger of an even greater overestimation of any stenosis.

1.4.1.3.2. 3D TOF sequence

In 3D acquisition, the whole of the stack of slices is acquired at the same time.

Figure 1.29. Optimal relative positions of the slices acquired in 3D TOF imaging and of the vessel
Adantages

– Better spatial resolution than in 2D mode: good visualization of small vessels.
– Good SNR.
– Good visualization of vessels containing areas of physiological turbulence (such as bifurcations) in comparison to 2D acquisition.

NOTE.– the problem of overestimation of stenoses, although it is less than with 2D TOF.

Disadvantages

– Longer acquisition times.
– Because of the excitation of a volume at each repetition, there is a progressive saturation of the vascular protons on the last slices downstream of the vessels, all the more so if they are slow-moving and the volume selected is thick. Thus, there is a reduction in the diameter of the vessels (see section 1.2.1.2.3). The slowest flows may even disappear completely. This type of acquisition is therefore not appropriate for the imaging of very slow flows.

Figure 1.30. Attenuation of the signal from the flow because of saturation in 3D TOF. A high level of gray indicates a weak signal

Solutions

In order to deal with this problem, we can decrease the saturation of the flows as they flow through the target volume by:

– Dividing the 3D acquisition into multiple blocks (or slabs).

These sequences are referred to as MOTSA (Multiple Overlapping Thin Slab Acquisition) sequences.
This technique has the advantage of combining good resolution (comparable to 3D imaging) whilst considerably reducing the saturation of the moving protons.

Figure 1.31. Comparison of attenuation of flow signals between in a “conventional” 3D sequence (left) and in a MOTSA sequence (right)

**NOTE.** It is necessary to cross-check adjacent volumes (overlapping) in order to prevent artifacts due to imperfections in the slice profile (the angle of excitation at the edge of the slice is less than at the center). If this overlapping is not done, we see the emergence of an artifact known as a “Venetian blind” effect.

Figure 1.32. Venetian blind artifact

Using a variable excitation angle, smaller at the entry of the flow into the volume and greater near the exit from the volume (*TONE: Tilted Optimized Nonsaturating Excitation*), we can reduce the progressive saturation of the vessels (this also enables us to decrease the signal from the tissues with a short $T_1$, thereby obtaining better contrast).

The choice of increment of the flip angle depends on the direction and velocity of the flow and on the thickness of the acquisition volume.
1.4.1.3.3. 2D / 3D “match”

<table>
<thead>
<tr>
<th>2D TOF</th>
<th>3D TOF</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optimal paradoxical enhancement</td>
<td>Non-optimal paradoxical enhancement: progressive saturation of the vessels.</td>
</tr>
<tr>
<td>Choice of orientation of slices, T₁ and flip angle</td>
<td>Solutions: TONE, MOTSA</td>
</tr>
<tr>
<td>Good contrast (because of good saturation of stationary tissues)</td>
<td>Poorer contrast</td>
</tr>
<tr>
<td>Sensitive to slow and fast flow</td>
<td>Sensitive mainly to fast flow</td>
</tr>
<tr>
<td>Poor spatial resolution: mediocre MIP reconstruction</td>
<td>High spatial resolution: good MIP reconstruction</td>
</tr>
<tr>
<td>Good visualization of large vessels with unidirectional flow</td>
<td>Good visualization of small vessels</td>
</tr>
<tr>
<td>Low SNR</td>
<td>High SNR</td>
</tr>
<tr>
<td>Short acquisition time</td>
<td>Long acquisition time</td>
</tr>
<tr>
<td>Very sensitive to turbulent flows (significant overestimation of stenoses)</td>
<td>Less sensitive to complex and turbulent flows (slight overestimation of stenoses)</td>
</tr>
<tr>
<td>Solution: Flow compensation gradients</td>
<td></td>
</tr>
<tr>
<td>Poor elimination of tissues with a short T₁ (fat, thrombi, hematomas, etc.)</td>
<td></td>
</tr>
<tr>
<td>Solution for hematomas or thrombi: phase-contrast MRA (PC-MRA)</td>
<td></td>
</tr>
</tbody>
</table>
1.4.2. Phase-contrast angiography (PCA)

1.4.2.1. Technique

1.4.2.1.1. Principle

Phase-contrast angiography is based on dephasing of moving spins subjected to a bipolar gradient.

Remember that in a bipolar gradient, the dephasing that the spins will experience when they move along the axis of the gradient is given by: \( \varphi = -\gamma \cdot v \cdot A \cdot \tau \)

where \( v \) is the velocity of the spins in the direction of the gradient, \( A \) and \( \tau \) are respectively the amplitude and time separating the two lobes of the gradient.

The sequences used are gradient-echo-type sequences.

We use a double acquisition:

- An acquisition with a first bipolar gradient (+A / -A )

\[ \begin{array}{c}
\text{tau'} \\
\end{array} \]

- For stationary protons, the phase obtained is equal to \( \varphi_{\text{static}} = \varphi_B \), with \( \varphi_B \) representing the phase due to inhomogeneities.

- For moving protons, the phase obtained is: \( \varphi_{\text{moving}} = -\gamma \cdot v \cdot A \cdot \tau + \varphi_B \)

- An acquisition with a second bipolar gradient, of inverse polarity to the first (+A / -A ).

\[ \begin{array}{c}
\text{tau'} \\
\end{array} \]

- For stationary protons, the phase obtained is equal to: \( \varphi_{\text{static}} = \varphi_B \)

The phase due to the inhomogeneities in the field is the same as for the first gradient (same environment for the stationary protons during the application of both gradients).
For moving protons, the phase obtained is: \( \Phi_{\text{moving}}^2 = +\gamma v \cdot A \cdot \tau + \Phi_B \)

Then, on a pixel-by-pixel basis, we subtract the signal obtained with the second gradient from the signal obtained with the first.

For stationary protons, the signal obtained is thus equal to:

\[
S_{\text{static}} = M \cdot e^{i\Phi_{\text{static}}} - M \cdot e^{i\Phi_{\text{moving}}} = 0
\]

For moving protons, the phase obtained is:

\[
S_{\text{moving}} = M \cdot e^{i\Phi_{\text{moving}}} - M \cdot e^{i\Phi_{\text{moving}}} = 2M \cdot e^{i\Phi_B} \cdot \sin(\gamma v \cdot A \cdot \tau)
\]

Conclusions

- This subtraction eliminates the background from the image (i.e. the stationary tissues for which the signal is null) and gives us an image where the signal is sensitive to the velocity \( V \) of spins moving in the direction of the gradient: this treatment is known as velocity coding.

- This coding is made possible even in the presence of inhomogeneities of the magnetic field (\( \Phi_B \neq 0 \)); this demonstrates the advantage to performing two acquisitions instead of one.

- In cases where a phase term due to the Foucault current appears, this method is no longer valid. Indeed, the phases acquired due to these currents in the first and second gradients (\( \Phi_e \) and \( \Phi'_e \)) are different: with the pixel-by-pixel subtraction, the signal from the stationary tissues will no longer be eliminated.

- The only option, in this case, to prevent disruption by these currents, is an appropriate antenna setup.

1.4.2.1.2. Method

In practical terms, if we wish to study the different flows in all directions in space, the method set out above requires six measurements (two measurements in each direction of flow coding), which is a relatively long process.

At present (thanks to the knowledge obtained on velocity-insensitive forms of gradient – see section 1.2.2.3), in practice the technique is modified, so that only four sequences are acquired:

- we perform one acquisition with flow encoding gradients in each of the three spatial directions (i.e. 3 acquisitions): we obtain the signals \( s_x, s_y \) and \( s_z \).
we perform an additional acquisition with a flow compensation gradient (i.e. a flow-insensitive acquisition) to obtain the reference signal $S_{\text{ref}}$.

Once the acquisitions have been made, we may then wish either to obtain the vascular anatomy, or perform measurements of velocity (and therefore flowrate), which do not involve the same data-processing procedures, as we shall now see.

**Imaging of vascular anatomy (angio-MRI)**

In the same manner as in i) above, we perform subtraction of the signals for each of the directions: $S_x - S_{\text{ref}} ; S_y - S_{\text{ref}} ; S_z - S_{\text{ref}}$.

These various complex subtractions are null for stationary protons and non-null for protons moving in the directions under examination.

We add together these three complex signals, and from the result we take the modulus:

$$S_{\text{anat}} = \left| (S_x - S_{\text{ref}}) + (S_y - S_{\text{ref}}) + (S_z - S_{\text{ref}}) \right|$$

$S_{\text{anat}}$ is non-null if there is a flow, regardless of its direction.

To create a 3D venous map, $S_{\text{anat}}$ is analyzed with an MIP algorithm.

**NOTE.**—It is of course possible to take, for instance, the modulus $|S_x - S_{\text{ref}}|$ without worrying about the other directions.

**Measuring the velocity (quantifying the flow)**

This time, it is the phases of the signals rather than the signals themselves that are subtracted, again for each of the directions:

$$\phi(S_x) - \phi(S_{\text{ref}}), \quad \phi(S_y) - \phi(S_{\text{ref}}), \quad \phi(S_z) - \phi(S_{\text{ref}})$$

We can then obtain a vector proportional to the velocity vector:

$$\overrightarrow{S_{\text{velocity}}} = \left[ \phi(S_x) - \phi(S_{\text{ref}}) \right] \hat{i} + \left[ \phi(S_y) - \phi(S_{\text{ref}}) \right] \hat{j} + \left[ \phi(S_z) - \phi(S_{\text{ref}}) \right] \hat{k}$$

$$\overrightarrow{S_{\text{velocity}}} = -\gamma \cdot A_x \cdot \tau_x \cdot v_x \cdot \hat{i} - \gamma \cdot A_y \cdot \tau_y \cdot v_y \cdot \hat{j} - \gamma \cdot A_z \cdot \tau_z \cdot v_z \cdot \hat{k}$$
NOTE.— As we saw in section 1.4.2.1.1.), the phase due to any inhomogeneities in the field is eliminated by subtraction.

We can thus establish a map with grayscale coding representing the norm of velocity at each point, and its orientation: flows running towards the tester are coded in shades of black (phase between -180° and 0°), and those running away from the probe are coded in shades of white (phase between 0° and 180°).

![Figure 1.34. Images of two flows of opposing directions in phase-contrast imaging](image)

Generally, outside of blood vessels, the image has a series of black and white dots representing background noise: when the two initial phases are subtracted (e.g. \( \phi (S_v) \) and \( \phi (S_{ref}) \)), as the noise factors associated with each of them are not correlated, the signal \( S_{velocity} \) has low intensity and a random sign outside of the vessels. This background noise can be eliminated if a threshold level for the value of the phase is imposed on the final image: the background then appears gray (i.e. an intermediary color).

![Figure 1.35. Phase-contrast imaging. Top: sagittal slice of location. Bottom: 2D PCA done in the slice marked by a line on the location image, perpendicular to the carotid and basilar arteries. 1 and 2: carotid arteries, and 3: basilar artery](image)
1.4.2.2. Choice of parameters and artifacts relating to the sequence

One of the main parameters needing to be determined in phase-contrast MRA is the velocity encoding $V_{\text{enc}}$. This is determined by the area $A \cdot \tau$ of the bipolar gradient, and is defined by the velocity of the protons giving a dephasing of 180° ($\pi$ radians):

$$\pi = -\gamma V_{\text{enc}} A \cdot \tau$$

The choice of velocity encoding is of crucial importance; if the velocity encoding is not at least equal to or greater than the maximum velocity of the protons in the vessels that we wish to image, a phenomenon of aliasing of the velocities occurs.

Hence, let us consider an velocity encoding equal to 50 cm/s; the protons circulating at this velocity have a phase of 180°.

Fast-moving protons, e.g. moving at 75 cm/s, will exhibit 270° dephasing.

Slow-moving protons, e.g. moving at 25 cm/s, and in the opposite orientation to the previous set of protons, will have -90° dephasing.

These two types of protons cannot be differentiated from one another, and are therefore coding with the same level of gray: this is the phenomenon of aliasing.

![Figure 1.36. Phenomenon of aliasing in PCA flow imaging](image-url)
The characteristics of the encoding gradient are therefore defined in order to be able to encode flows within a certain range of velocity: between \(-V_{\text{enc}}\) and \(+V_{\text{enc}}\). Any velocity beyond this range will be incorrectly encoded.

In order to avoid aliasing, we need to estimate the highest velocities in the vessels which we wish to image.

In practice, velocity aliasing is not massively problematic, because the flow velocities are usually greater in the center of the vessel (this is known as a laminar flow): the vessel will be clearly delimited by the protons circulating at a lower velocity at the edges.

Before launching a sequence, we need to test various velocity encodings in order to choose the one which is most appropriate (this is generally done with a 2D PCA sequence: see below); it should be noted that the velocity encoding may be different in each spatial direction.

**Example**

A velocity encoding of 20 to 40 cm/s is indeed appropriate for venous flow.
A velocity encoding of 60 to 80 cm/s is appropriate to visualize the cerebral arteries (to the detriment of slow flows such as venous flow, whose signal would be very weak).

Other parameters relating to the method

Similarly to TOF-MRA, the $T_R$ is chosen to be short (30-100ms), but care is taken not to cause the saturation of the vessels in the slice of interest.

The limitations of this technique relate to the loss of signal caused by complex or turbulent flows that is responsible for intravoxel dephasings: in order to reduce this artifact, the echo time $T_E$ will be chosen as short as possible (8-14ms).

This loss of signal is all the greater when the voxel is large; in 3D imaging, therefore, we always try to choose a small voxel size.

Finally, the artifact of MIP and placement of the acquisition volume, which we saw above in our discussion of TOF sequences, is also present with the PCA sequence.

1.4.2.3. Different types of acquisition of a volume

1.4.2.3.1. 2D PCA sequence

The advantages to 2D PCA are several:

– its rapidity (a slice can be acquired in under two minutes), which means we can test different encoding velocities for a possible 3D acquisition;

– the possibility of creating cinematic imaging of the vascular flows (with ECG synchronization);

– the option to plot a curve showing the velocity over time within the same slice. This velocity curve is then coupled with a morphological image to give us the breadth of the vessels and deduce the vascular flowrate.

1.4.2.3.2. 3D PCA sequence

The main advantages to 3D PCA over 2D PCA are similar to those of 3D TOF over 2D TOF:

– much better resolution, so less sensitive to turbulent flows;

– improved SNR.

The main disadvantage of the sequence is its relatively long acquisition time (over 10 minutes); this time is twice as great as for 3D TOF imaging (because twice
as many acquisitions are made). In an attempt to reduce it, we can decrease the number of phase encoding steps, or use parallel imaging (SENSE).

<table>
<thead>
<tr>
<th>2D PCA</th>
<th>3D PCA</th>
</tr>
</thead>
<tbody>
<tr>
<td>– Very good elimination of stationary tissues (even those with short T&lt;sub&gt;1&lt;/sub&gt; times)</td>
<td>– Poor spatial resolution</td>
</tr>
<tr>
<td>– Very good contrast</td>
<td>– Better spatial resolution</td>
</tr>
<tr>
<td>– No MIP reconstruction</td>
<td>– Good quality MIP reconstruction</td>
</tr>
<tr>
<td>– Sensitive to slow and fast flows</td>
<td>– Very sensitive to complex and turbulent flows (overestimation of stenoses)</td>
</tr>
<tr>
<td>– Short acquisition time</td>
<td>– Very long acquisition times</td>
</tr>
<tr>
<td>– Very sensitive to complex and turbulent flows (overestimation of stenoses)</td>
<td>– Less sensitive to complex and turbulent flows (less than 2D PCA but more than 3D TOF) (slight overestimation of stenoses)</td>
</tr>
</tbody>
</table>

1.4.3. The “match” between TOF and PCA imaging

The table below illustrates the advantages of one method over another.

<table>
<thead>
<tr>
<th>TOF</th>
<th>PCA</th>
</tr>
</thead>
<tbody>
<tr>
<td>– Appropriate for vessels with slow to fast flow</td>
<td>– Appropriate for vessels with very slow to fast flow</td>
</tr>
<tr>
<td>– Better visualization of the vessels containing physiological turbulences</td>
<td></td>
</tr>
<tr>
<td>– Shorter acquisition times with identical spatial resolution</td>
<td></td>
</tr>
<tr>
<td>– Better spatial resolution with identical acquisition times</td>
<td></td>
</tr>
<tr>
<td>– Better contrast because of better elimination of stationary tissues</td>
<td></td>
</tr>
</tbody>
</table>

Conclusion to this table: the choice between TOF and PCA is dictated by the region needing to be explored and the type of bloodflow: 3D TOF offers the best performance for the exploration of intracranial vessels (good resolution needed, fast flow); 2D- or 3D PCA sequences are better adapted for studying the cerebral veins (slow flows).
1.4.4. Contrast-enhanced MRA (CE-MRA)

With the two techniques discussed above (TOF and PCA), MRA has taken off and has become part of medical routine, particularly in the study of the brain.

However, there are three main criticisms that can be leveled at these techniques:

– the acquisition times are particularly long;

– the issues of the protons’ saturation mean that the volumes which can be explored are small (i.e. limited FOVs, fields of view);

– the temporal resolution obtained with these methods is very poor (there is no possibility of dynamic tracking of the flow).

With a view to getting around these difficulties, a new technique has been developed: CE-MRA.

1.4.4.1. Technique

1.4.4.1.1. Principle and advantages of the sequence

CE-MRA is an ultra-fast 3D gradient-echo sequence with a spoiler gradient placed after the acquisition to counter the residual magnetization and obtain significant weighting $T_1$.

The technique involves the injection of a paramagnetic substance: gadolinium (Gd). The gadolinium creates an intense local magnetic field which disturbs the environment of the protons of the medium in which it is: thus, their relaxation is accelerated, and the Gd artificially lessens the $T_1$ of the blood (and the $T_2$, but only with high doses).

![Graph showing $T_1$ and relaxation of blood (with and without gadolinium) and fat in a field $B_0 = 1.5 T$](image)

**Figure 1.38.** Compared values for $T_1$ and relaxations of blood (with and without gadolinium) and fat in a field $B_0 = 1.5 T$.
Thus, the vascular signal obtained with this sequence is intense, and relatively independent of the flow artifacts discussed above:

– no saturation of the blood signal, which enables us to explore large volumes;
– little loss of signal relating to intravoxel dephasing, which means we can image turbulent flows.

**NOTE.**– There is always intravoxel dephasing, but the consequent decrease in the $T_1$ “masks” the effects of this dephasing.

In addition, CE-MRA is set apart from PCA and TOF sequences by virtue of the fact that:

– it is faster: the acquisition is done in the space of a few seconds;
– it provides better contrast: the stationary protons are saturated thanks to the short $T_R$ times;
– it decreases the motion-related artifacts: there is the option of performing sequences under controlled apnea (so there is less sensitivity to these artifacts).

This form of acquisition, after treatment (MIP algorithm, as used with TOF or PCA sequences), yields images similar to those obtained by “conventional” angiography, with the advantage of not being irradiant.

**Figure 1.39.** CE-MRA of the abdominal aorta and its renal branches. *(From [AUE 04]. Reproduced with kind permission from Elsevier)*

1.4.4.1.2. Description of the k-space and injection of gadolinium

We know that the central rows in the k-space play a part in the contrast of the image and the SNR, whereas the peripheral rows determine the spatial resolution.
Thus, in order to obtain as high a contrast as possible, the center of the k-space needs to be recorded when the concentration of gadolinium (when it first passes through) in the slice is maximal (maximal signal).

This concentration peak is very fleeting, so we need to:

- **choose a good definition of the k-space** which favors a quick start to acquisition, enabling the center of the space to be acquired rapidly:
  - row-by-row definition (in a linear or sequential fashion). Creation of the contrast of the image requires 20% of the acquisition time, Figure 1.40. *Cartesian definition of k-space*

  - elliptical-central definition: the center of the k-space is obtained first by creating an ellipse, which is equivalent to acquiring the data by a range of frequencies (from the lowest to the highest). Creation of the contrast of the image now requires no more than 2% of the acquisition time. Yet the image quality may be worse (overall blur) than with row-by-row scanning if the gradients are not properly stabilized; Figure 1.41. *Elliptical-central definition of k-space*

  - know the transit time of the gadolinium from the injection point to the vessels being imaged. This time corresponds to an acquisition window between 5 and 20 seconds (for the slowest flows).
A variety of methods are available to perform good synchronization:

– carry out a test injection and acquire a slice centered on the region of interest every second for a minute. A region of interest (ROI) enables us to determine the transit time by pinpointing the peak of the signal. This method is relatively slow but it is the most reliable;

– perform automatic detection: a volume of interest (VOI or tracker) is positioned upstream of the vessel needing to be explored;

– a TSE (Turbo Spin Echo) sequence automatically analyzes the intensity of the signal in the vessel. When the threshold is reached, the CE-MRA sequence is triggered, either automatically or after a fixed delay;

– the main drawback to this method is the danger of incorrect analysis of the intensity of the signal, polluted by motion-related artifacts, for instance;

– semi-automatic detection: real-time visualization of the arrival of the gadolinium in the vessels with an ultra-fast (1FPS) 2D gradient-echo dynamic sequence.

1.4.4.2. Parameters relating to the sequence

All the usual criteria used in a “conventional” sequence need to be reviewed, bearing in mind that we want to make a 3D acquisition very fast with a wide FOV.

1.4.4.2.1. Repetition time, echo time and sweep width

These two times are chosen to be as short as possible so as to decrease the total acquisition time: $TR \leq 5\, ms$ and $TE \leq 2\, ms$.

It is not helpful to compensate the flow (see section 1.2.2.3): indeed, the gain in $T_1$ masks all the flow-related phenomena. Furthermore, any compensation would prolong the $TE$.

**NOTE.**– With a CE-MRA sequence, it is possible to decrease the echo time. Indeed:

![Diagram](image)
It is also possible to use specific echo times (2.2 ms at 1.5 T) where the fat and water are either in phase or in phase opposition, which enables us to remove the image of the fat.

1.4.4.2.2. FOV

The choice of the orientation of the volume to be imaged is entirely independent of the orientation of the vessels (unlike TOF and PCA) because the phenomenon of saturation is no longer present.

With CE-MRA, it is possible to obtain extended FOVs: thus, we can image the abdomen, the lower limbs, etc.

A compromise

If the FOV is widened and the size of the matrix remains the same, the spatial resolution suffers; if we wish to preserve the same spatial resolution, it is necessary to increase the acquisition time.

A compromise needs to be found between acquisition time, FOV, spatial resolution and SNR.

Solutions

– Generally, the FOV and the matrix are rectangular (as they are in “classic” imaging).

– We can perform zero-filling in the direction of the slice selection gradient (which reduces the acquisition time if the patient moves too much, for instance).

– Even with the two solutions suggested above, the acquisition time remains relatively long and the spatial resolution is sub-optimal; this is true even with the best technical characteristics of the gradients (which determine the acquisition times).

Parallel acquisition techniques, which deliberately cause aliasing of the image (in the k-space with SMASH and in the real domain with SENSE) help to considerably reduce the acquisition time or significantly increase the spatial resolution without placing further constraints on the gradients.

In addition, CE-MRA has an intrinsic SNR which is sufficient to compensate for the loss due to the use of these techniques.

*Supra-aortic angio-MRI uses these techniques. Venous angio-MRI does not, as time-gain is not crucially important in this application.*
Generally, the spatial resolution in CE-MRA is poorer than that achieved with 3D TOF for studying intracranial vessels.

1.4.4.2.3. Calculating the injection delay

The passage of the gadolinium through the slice needs to coincide with the acquisition of the central rows in the Fourier plot. A precise timing therefore needs to be established, giving the delay $T_{dem}$ between the injection and the start of acquisition.

For this purpose, it is useful to describe the evolution of the signal over time that is characteristic of its evolution.

![Diagram showing signal intensity evolution and injection timing](image)

**Figure 1.42. Signal intensity in the slice of interest and injection of gadolinium**

*Criteria and parameters needing to be respected*

The duration of infusion of the bolus $T_{inj}$ (i.e. the time period over which the gadolinium injection is administered) needs to cover at least half the acquisition time $T_{acq}$ so that the central rows of the k-space can be acquired with maximum signal strength:
– the transit time $T_{\text{transit}}$ can vary between 6 and 40 seconds: this time can be evaluated with an injection test;

– generally, in order to improve the visibility of the images obtained, we need to choose between arterial and venous imaging.

This choice can be made on the basis of the period when the acquisition needs to take place: if we wish to avoid venous contamination, the acquisition needs to be made during the first phase of the gadolinium’s arterial passage.

In the case of cerebral imaging, for instance, the acquisition time $T_{\text{acq}}$ needs to be very short (10-12 seconds) which is the approximate length of time taken for blood to complete the cerebral arterial circuit.

Given these different times, referring to the curve giving the intensity of the signal in the slice over time, we have:

$$T_{\text{dem}} = T_{\text{transit}} + \frac{T_{\text{inj}}}{2} - \frac{T_{\text{acq}}}{2}$$

1.4.4.2.4. Flip angle

There is less of an impact on the vascular contrast than with PCA and especially TOF. Habitually, the flip angle used for arterial imaging is between 50° and 60°, and for venous imaging between 30° and 40°.

---

**Figure 1.43.** CE-MRA showing a bilateral occlusion of the superficial femoral artery.  
(From [AUE 04]. Reproduced with kind permission from Elsevier)
1.4.4.3. **Contrast-enhancing strategies**

1.4.4.3.1. Before acquisition of the sequence

Preparatory sequences (STIR or FLAIR) can be integrated with the CE-MRA sequence. These help to reduce the signal from the stationary tissues, although they do not eliminate it entirely.

On the other hand, they do lead to a consequent increase in acquisition time.

1.4.4.3.2. After acquisition of the sequence

The taking of an acquisition before the bolus arrives (see Figure 1.38) can be used to perform a *subtraction*, as done with the PCA sequence, to eliminate the stationary tissues, including those with short $T_1$ times.

This subtraction is crucial in order to view small structures such as the intracranial vessels.

However, as it requires the patient to remain perfectly still, subtraction is not used in regions where the movements (heart pulses, breathing movements, etc.) are significant, such as in the neck.

1.4.4.4. **Artifacts**

1.4.4.4.1. Artifact of timing

As we saw above, the speed of the CE-MRA sequence and the fleeting passage of the gadolinium through the slice of interest necessitate very precise timing.

If this condition is not respected to the letter, certain artifacts may arise.

In particular, when the acquisition is triggered too late, a venous signal appears which “pollutes” the image, rendering it less clear.

1.4.4.4.2. Artifact of MIP and placement of the acquisition volume

This artifact has already been discussed with regard to the TOF and PCA methods, but is also observable with CE-MRA.
1.4.4.5. The positives (+) and negatives (–) of the CE-MRA technique

<table>
<thead>
<tr>
<th>+</th>
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<tbody>
<tr>
<td>• Wide FOV</td>
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<tr>
<td>• Excellent elimination of stationary tissues (subtraction technique)</td>
</tr>
<tr>
<td>• Short acquisition time (20-40s)</td>
</tr>
<tr>
<td>• Less sensitivity to movements</td>
</tr>
<tr>
<td>• Fairly good temporal resolution (but greatly inferior to that achieved with “conventional” angiography, which is 0.5 s)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>–</th>
</tr>
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<tbody>
<tr>
<td>• Independent of flow-related phenomena</td>
</tr>
<tr>
<td>• Good spatial resolution (little intravoxel dephasing)</td>
</tr>
<tr>
<td>• Not hugely sensitive to turbulent flows</td>
</tr>
</tbody>
</table>

• More invasive
• Higher cost
• Negative effect on the $T_2^*$ of the gadolinium at high doses

NOTE.– With the aim of further improving the temporal resolution of CE-MRA, it is possible to perform dynamic angio-MRI: the 3D sequences obtained are optimized and the temporal resolution is around 1.4 s, which is still less than that obtained with conventional angiography.

1.5. Study of two clinical problems

The choice of which type of sequence to use to image the flow, from amongst the various options advanced in this chapter, must be based on the type of disease from which it is suspected that the patient is suffering.

Indeed, if we choose an “incorrect” sequence, there is a danger of causing confusion and artificially aggravating the problems detected.

1.5.1. Differentiation between a thrombus and a slow flow

1.5.1.1. The problem

The distinction between a thrombus (a mass of coagulated blood that is formed in a vessel) and a slow flow is a problem that is regularly encountered in clinical routine.
In the case of a conventional TOF sequence, the signals from a thrombus and a slow flow are seemingly the same, because:

– the slow flow experiences a slight phenomenon of paradoxical enhancement;
– the thrombus has a fairly short $T_1$ and a relatively high $T_2^*$ (owing to the presence of methemoglobin, which has a paramagnetic effect).

How, in this case, can we differentiate a thrombus from a slow flow?

1.5.1.2. Proposals and solutions

A number of strategies are put forward below. For each of them, we indicate whether it could indeed be a solution to the problem at hand, and the reasons for this judgment.

– Increasing the TE? $\rightarrow$ YES

This increase in TE leads to a significant weighting $T_2$ (or $T_2^*$) of the MRA sequences:

- the thrombus has a relatively strong signal regardless of the value of the TE;
- the signal from the slow flow will fade.

– Observation of artifacts? $\rightarrow$ YES

If a 3D acquisition is performed, is there a *decrease in the signal in the direction of the flow* within the volume? If so, we are witnessing a slow flow.

Is there an *artifact of ghosting* near to the vessel? If so, we are looking at a slow flow.

– Injection of gadolinium? $\rightarrow$ NO

This injection cannot solve the problem, because the signals from a slow flow and from a thrombus will both be increased.

– Application of a pre-saturation slab? $\rightarrow$ YES

This strip needs to be placed above or below the structure that is suspected of being a thrombus, in order to validate or debunk this hypothesis: if the signal is reduced, then we are dealing with a slow flow.

– Use of PCA? $\rightarrow$ YES and NO

This is possible if the velocity encoding is sufficiently low to enable us to see slow flows. The signal from the thrombus still should not appear. However, take care: extremely slow flows will still not appear.
– Two acquisitions with TOF sequences?  ⇒ YES and NO

As we saw above, a single TOF sequence is not enough. Two TOF sequences might alleviate the doubt if:

- those two sequences are acquired with slices oriented in different directions;
- one of the two sequences has a flow compensation gradient and the other does not.

In both cases, the signal from a slow flow will be altered, but the signal from a thrombus will not.

1.5.2. “Correct” evaluation of a stenosis

1.5.2.1. Problem

The correct evaluation of the diameter of a vessel is made difficult by the loss of signal at the periphery of that vessel engendered by the phase heterogeneity of the protons within the same voxel (see section 1.3.3.1).

In order to deal with this problem, we saw in our discussion of flows at constant velocity that it was possible to use a sequence with flow compensation gradients.

The case of stenosis (constriction of the caliber of a vessel) is different: it is not possible to rephase the signal with a flow compensation gradient because the acceleration of the blood is non-null.

1.5.2.2. Possible solution

The above problem stems from the fact that the sequences used (with a flow compensation gradient) show flow with constant velocity as a hypersignal and stationary tissues as a hyposignal.

Hence, exhibiting acceleration is confused with the stationary tissues.

One possible solution is to image the stationary tissues as a hypersignal and “the rest” (flow with any velocity, be it constant or otherwise) as a hyposignal.

The sequence used for this type of imaging is a black-blood TOF sequence.