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Atlas of Practical Applications of Cardiovascular Magnetic Resonance

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Springer
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Six years have passed since the edition of our Atlas of Practical Cardiac Applications of MRI. Fortunately, the technique has experienced during this time a continuous development that demanded a new updated version of the book. One of the consequences of this growing process has been the adoption of the term Cardiovascular Magnetic Resonance (CMR) to refer to the technique, and this is why the title of the present version of the book has changed slightly in relation to the first one.

Merits of CMR were evident from its beginning, early in the eighties. There was a time, however, when CMR ran the risk of becoming a luxury diagnostic tool, either confined to the experimental field or limited to serve as an occasional resource for selected clinical entities, only applied by even more selected specialists. Things have changed, for the benefit of cardiology, during the last few years, particularly since CMR has proved to be a very useful technique also in the study of ischemic heart disease, which constitutes the main issue of concern in cardiology today. As a consequence, CMR is starting to play a relevant role in clinical practice, and it has become an issue of interest for the educated cardiologist. Evidence for this is the increasing body of research articles which, on CMR in myocardial infarction, for instance, has more than doubled since 1999 compared with the five-year previous period, or the publication of major textbooks on CMR, up to three in the last two years. Also, there is a strong academic interest on the technique, with a growing number of specialized meetings and activities from associations aimed to spread knowledge on the technique, as the Society for Cardiovascular Magnetic Resonance, or the Working Group on CMR of the European Society of Cardiology.

Updated sources of information are thus needed for clinicians interested in the field and, particularly, for those demanding to be initiated in it. The present version of this Atlas is intended to be one of these sources. Most of the authors are clinical cardiologists devoted to the field of cardiac imaging who have the privilege of being experienced in those techniques that are ultimately useful on the clinical problems they are dealing with. During the last 25 years, we have learnt, as every cardiologist involved in cardiac imaging has done, that the actual value of a technique lies in its ability to give reliable answers to relevant clinical questions. When such a technique is, in addition, unique in doing that, then it becomes essential. Echocardiography has fulfilled these requirements in many aspects for many years, and its role is not disputed. CMR is gaining the same respect also in a good deal of issues, and today is no longer an auxiliary tool but a first line method in many instances. The important point is to learn which ones. We hope that the examples presented in this book will help clinicians to learn when a CMR exam is useful and, also, how to plan and read CMR studies in every particular case.

The emergence of radiological techniques, like CMR or computed tomography, in the field of diagnostic cardiology, has raised the question of which specialist should be in charge of these studies. The technical complexity of CMR, and the wide variety of pulse sequences and acquisition strategies, made reasonable the direct participation of radiological teams. On the other hand, the even more complex spectrum of cardiovascular disorders and, particularly, the different clinical relevance of some of their aspects, seem to demand a cardiological view. In practice, excellent teams performing CMR may be found among radiologists specialized in cardiovascular diseases and, also, among cardiologists devoted to imaging, a close cooperation between them being the optimal approach. In our case, we have always agreed with our radiologist colleagues on the collaborative nature of this task. We are in debt to them, and we feel mandatory a mention of the radiology departments where the studies illustrating this book have been obtained:

- Radiology Department (Dr. Jordi Ruscalada), Hospital de Santa Creu i Sant Pau, Barcelona, Spain.
- Corachán Magnetic Resonance Center (Dr. Carlos Alexander), Clínica Corachán, Barcelona, Spain.
- Balearic Magnetic Resonance Center (Dr. Dario Taboada), Clínica Femenia, Palma de Mallorca, Balearic Islands, Spain.
- Clínica Ruber and Hospital Ruber Internacional, Madrid, Spain

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Axial plane: horizontal transverse plane.

Black Blood Imaging: MR sequence in which blood does not generate signal, such as spin-echo sequences.

Breath-hold: voluntary interruption of the respiration necessary for the correct acquisition of fast sequences.

Cine MR: series of gradient-echo images obtained in consecutive phases of the cardiac cycle and displayed in a continuous loop sequence.

Coil: element of the MR system that generates the radiofrequency pulses used to excite the study subject (transmitting antenna), or that also receives the echoed pulses returned by tissues (receiving antenna). The same coil can act as both transmitter and receiver, or there can be two independent coils, one for each function.

Contrast: The difference in signal intensity between tissues in an image.

Contrast agent: A substance administered to the patient in order to change the contrast between tissues (gadolinium-based contrast agents are used in clinical practice).

Coronal plane: vertical frontal plane.

Delayed contrast-enhanced imaging: CMR method that, by using Inversion-Recovery sequences, allows the identification of myocardial scar tissue due to the persistence of the contrast agent in areas of myocardial fibrosis up to 30 minutes after its administration, while it is washed-out from the normal myocardium, this resulting in a high signal intensity of the scarred tissue.

Echo planar imaging (EPI): technique enabling the obtention of an image by means of a single radiofrequency excitation in a time on the order of milliseconds.

Echo time (TE): time interval between the radiofrequency pulse emission excitation and the reception of the radiofrequency signal emitted by the tissues.

Fast low-angle shot: gradient-echo sequence that uses short repetition times and reduced matrix, thus allowing the acquisition of images in less than a second.

Fast SE sequence: A multiple echo spin-echo sequence which allows the acquisition of several lines of k-space within a repetition time.

Field strength: degree of intensity of the magnetic field generated by the system magnet (measured in Tesla units).

Field of view (FOV): dimension of the study window.

Flip angle: value reached by the precession angle when stimulated by a specific radiofrequency pulse.

Free induction decay (FID): name given to the radiofrequency signal emitted by the protons of tissue during relaxation after having been submitted to radiofrequency excitation at resonance frequency in the presence of an intense, external magnetic field.

Frequency encoding: a process which enables the location of a point along one of the axes of the study plane: along with phase encoding, it defines the position of this point in the study plane.

Gadolinium (chelated): paramagnetic contrast agent of intravascular and extracellular distribution that produces a change in T1 and T2 relaxation times of the tissues this improving their contrast in the images.

Gating: coupling between slice acquisition of a sequence and any cyclical physiological signal: ECG; respiration, peripheral pulse.

Gradient echo (GE): MR technique by which adequately contrasted images are obtained of dynamic structures and of blood flow. Due to the short repetition times employed, it is possible to include various excitations in one cardiac cycle time, which enables the obtention of dynamic cine MR sequences.

Interslice gap: distance, which does not appear in the image, that separates contiguous slices of a sequence.

Inversion Recovery sequence: MR sequence in which, by applying a 180° inversion RF pre-pulse, signal intensity and contrast between tissues is modified.
Inversion time: Time interval between the inversion pre-pulse and the acquisition of the echo in an Inversion-Recovery sequence.

K-space: Numerical matrix containing the information needed to produce an image. The Fourier transform (mathematical method) of k-space results in an image.

Magnetic resonance angiography (MRA): Contrast-enhanced MR technique that provides a 3D imaging of the vessels.

Matrix: number of information units (voxels) that constitute the image.

Multi-phase: any sequence in which each slice is obtained in multiple phases of the cardiac cycle.

Multi-slice: any sequence in which multiple contiguous slices are obtained during one acquisition.

Oblique plane: plane with a certain degree of angulation over any one of the standard planes (axial, coronal or sagittal).

Parallel imaging: Imaging technique that allows reconstruction of an image in less than the time required for conventional imaging by using the spatial information related to the different coils of the receiver array.

Phase encoding: a process which enables the location of a point along one of the axes of the study plane: along with frequency encoding, it defines the position of this point in the study plane.

Pixel size: dimensions (in mm) of the information unit in the two-dimensional representation of the image on the screen. Image resolution depends on pixel size, which varies according to field of view and matrix.

Precession angle: angle between the vector axis of the external magnetic field and the rotation axis of the hydrogen proton.

Pulse sequence: series of consecutive excitations and receptions, its analysis resulting in the acquisition of images with any one of the MR techniques.

Radiofrequency: fragment of the electromagnetic spectrum that includes waves with frequencies under $10^{12}$. The radiofrequency waves used in MR have frequencies of 10 to 100 MHz.

Radiofrequency pulse: brief radiofrequency signal emitted to excite the protons of the hydrogen atoms of the study subject.

Relaxation time: time required for the returning to a resting state of the hydrogen protons after a radiofrequency pulse excitation. Longitudinal relaxation or T1 is the time it takes them to return to the basal precession angle. Transversal relaxation time or T2 is the time elapsed until the energy acquired by phase coherence, in which protons are under the influence of an external radiofrequency pulse, is lost.

Repetition time (TR): interval between the emission of two radiofrequency pulses.

Sagittal plane: vertical antero-posterior plane.

Scout image: initial planes of rapid acquisition used to locate the next sequences.

Segmented-K-space: fast imaging technique based on the obtention of grouped lines of information (segments) instead of the line-by-line method used in conventional techniques, this reducing the acquisition time of an image to a matter of seconds.

Signal void: area of absent signal due to the flow characteristics in a specific region of the slice plane: it appears in instances of high flow rate or of turbulence in gradient-echo sequences.

Signal averaging or number of excitations (NEX): number of repeat measurements required for a sequence to be obtained with adequate definition of the images.

Signal-to-noise ratio (SNR): relation between the signal intensity from tissue structures and the background image noise, upon which image quality depends.

Single phase: any sequence in which one or various slices are obtained, each one in a different phase of the cardiac cycle.

Single slice: any sequence in which the images are obtained from a single slice.

Slice thickness: width of the slice.
Slice: section of the study subject which is imaged.

Spatial resolution: ability to discriminate between two different structures in the image, depending on the field of view and the matrix size.

Spectroscopy: technique that permits the acquisition, in an area of a specific tissue displayed in the MR image, of the spectrum of concentrations of an element (usually phosphorus) according to the different chemical compounds in which it is present.

Spin echo (SE): MR sequence that provides images of adequate contrast between tissues and blood flow, as no signal is elicited by rapidly moving structures.

Steady state free precession imaging (SSFP): GRF sequence that provides faster acquisition times with a higher contrast between tissues than conventional sequences.

Tagging: MR technique in which equidistant crossing lines are magnetically preselected in the ventricular myocardium, allowing to track dynamic myocardial wall changes during the cardiac cycle.

Tesla: standard unit of magnetic field strength.

T1-weighted image: MR image in which the signal intensity of the tissues is predominantly dependent on T1.

T2-weighted image: MR image in which the signal intensity of the tissues is predominantly dependent on T2.

Ultrafast sequences: techniques applying information acquisition strategies designed to reduce the total time spent in the process of imaging.

Velocity-encoded MR imaging: MR sequence which permits flow measurements through a particular vessel by providing both, the area and mean velocities of the blood flow, of the vessel studied.

Voxel: tridimensional unit of the MR matrix that integrates the two dimensions constituting the pixel with the thickness of the slice.

Wash-out effect: effect by which the blood flowing in a direction perpendicular to the slice plane produces a characteristic absence of signal (signal void), in the spin-echo technique, due to the fact that the excited blood leaves the slice plane well before the echo signal is read.
1.1 Definition and historical background

Magnetic Resonance (MR) is a physical phenomenon consisting on the emission of a radiofrequency (RF) signal by certain nuclei with an angular momentum (spin), such as hydrogen (\(^1\)H, proton), carbon (\(^13\)C), or phosphorus (\(^31\)P), after having been stimulated by RF pulses while being under the influence of an external magnetic field (B0). This phenomenon is based on the principle that these nuclei can absorb energy at a specific frequency, known as Larmor frequency.

Clinical MR imaging is based on the detection of hydrogen nuclei, which are widely present in all structures from the body in the form of water (H\(_2\)O) and fat. The physical phenomenon of nuclear MR was discovered in 1946\(^1\). The first clinical images obtained by this technique were published in 1973\(^2\). The evolution of computer technology in the 1980’s allowed the development of first commercial equipments, with most applications focusing on the brain. Some early reports of cardiac applications were published in 1983\(^3\). Since then, cardiovascular MR (CMR) has experienced a continuous expansion\(^4,5\) as is demonstrated by the amount of clinical reports on its applications, including reports from expert committees summarizing the present indications of the technique\(^6\). Currently, CMR is the newest and most complex of the cardiovascular imaging modalities, with diverse clinical applications spanning nearly every aspect of disease affecting the heart and blood vessels\(^7\). Noteworthy, 2003’s Nobel prize in Physiology or Medicine was awarded to Paul Lauterbur and Peter Mansfield, who were recognized for their contributions to the basis for the development of magnetic resonance into a useful imaging method in the early 1970s.

1.2 Physical basis

Nuclei with a spin, exposed to an external magnetic field, act like magnetic dipoles. Magnetic dipoles under the influence of an external magnetic field (B0) do not align in a strictly static position. Instead, they exhibit an
oscillation around their own axis parallel to the magnetic field (precession movement or spinning) (Figure 1.1). The dipol’s deviation from the axis of the external field is called the angle of precession. Hydrogen nuclei have only two possible orientations in a magnetic field: parallel (low energy) or anti-parallel (high energy) (Figure 1.2).

The precession frequency of these nuclei is called Larmor frequency. It is proportional to the external magnetic field, with the proportionality factor \( \gamma \) usually given in units of MHz/T. For protons, \( \gamma = 42.6 \) MHz/Tesla, resulting in a Larmor frequency of 63.9 MHz at 1.5 T, a typical field strength in clinical MRI.

1.2.1 MR signal: Free Induction Decay
In the presence of an exterior \( B_0 \) field, a slightly higher number of spins have a \( z \)-component parallel to \( B_0 \) (\( z \)-axis) as compared to anti-parallel. This polarization is called net-magnetization. However, since the spins are oriented randomly with respect to the other two axes (\( x \)– and \( y \)-axis), the integral magnetization is parallel to \( B_0 \). Therefore no precession occurs and no signal is detectable (Figure 1.3). When protons under the influence of an external magnetic field receive an energetic impulse by means of a RF wave pulse emitted at its resonance frequency, a deviation of the angle of precession is produced in the xy plane (flip angle) (Figure 1.4). Flip angle depends on the amplitude and duration of the RF pulse. On the other hand, the RF pulse induces all protons to precess in-phase, resulting in phase coherence. When the RF signal is interrupted protons are in phase. This oscillation can be detected by external receiver coils and results in a signal known as FID (Free Induction Decay) (Figure 1.5). However, the phase coherence will decay over time, and the spins will progressively and individually return to their basal state. The detection of this RF wave by a receiving coil constitutes the base for both the technique of MR imaging as well as for MR spectroscopy.

1.2.2 Relaxation
The return of the hydrogen nucleus to the basal state following the cessation of the RF pulse is called relaxation. Magnetization has vectorial characteristics and can be considered as constituted by a longitudinal component, parallel to the axis of the external magnetic field (\( z \)-axis), and by a transversal component orthogonal to it (\( xy \)-plane). The time constant, at which the longitudinal vector returns to its value prior to stimulation by the RF pulse is called longitudinal relaxation time, or \( T_1 \). Recovery occurs exponentially as shown in a graph of signal intensity against time in Figure 1.6 (A). \( T_1 \) depends on the tissue and on the external field strength. Typical values of \( T_1 \) of biological tissue are in the order of a few hundred ms to a few seconds under field strengths available for clinical applications (Table 1.1). Longitudinal relaxation time depends on the molecular environment: \( T_1 \) tends to be long in solids and short in mobile liquids. Differences in \( T_1 \) between different tissues are partly responsible for MRI contrast between tissues.

The transversal vector (in the \( xy \)-plane perpendicular to the axis of the external magnetic field) reflects the energy secondary to the coherence in the spin phase of the different protons. In the absence of RF stimulation, each hydrogen nucleus spins with a different phase. The resultant net value is thus zero. When an RF pulse is applied, the nuclei couple in a phase (all spinning together at the same time) and the transversal vector is then maximal. After the RF signal has been switched off, the coherence and, thus, the energy contained in it dissipates. The loss of energy due to dephasing coherence is known as transversal relaxation. It occurs as an exponential decay with a time constant \( T_2 \) (Figure 1.6 B). Transversal relaxation is conditioned by two factors: first, due to physical inhomogeneities in the local magnetic field, known as \( T_2^* \); and, second, by tissue composition (spin-spin interaction): \( T_2 \). \( T_2^* \) is a much faster process than \( T_2 \), but its effect can be reversed by applying a second RF signal pulse with a phase shift of 180°. This procedure eliminates the nonvariable components while retaining only the influence of the tissue composition in the signal. Transversal relaxation time is very sensitive to the presence and type of other atoms that surround hydrogen originating the resonance signal. Therefore, it varies greatly according to the different tissues, but it is practically unaffected by the intensity of the external magnetic field. Typical values
FIGURE 1.1

Precession movement or spinning

FIGURE 1.2

Parallel
Low energy

Anti-parallel
High energy

FIGURE 1.3

A.

B.

Table 1.1 T1 values at 1.5T

<table>
<thead>
<tr>
<th>Tissue</th>
<th>T1 (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Blood</td>
<td>1200</td>
</tr>
<tr>
<td>Myocardium</td>
<td>880</td>
</tr>
<tr>
<td>Muscle</td>
<td>880</td>
</tr>
<tr>
<td>Lung</td>
<td>820</td>
</tr>
<tr>
<td>Fat</td>
<td>260</td>
</tr>
</tbody>
</table>