Cardiovascular magnetic resonance imaging

Theodoros Karamitsos

Stefan Neubauer

Abstract

MRI (magnetic resonance imaging) uses the magnetic properties of the hydrogen nucleus, radio waves and powerful magnets to provide high-quality still and cine images of the cardiovascular system, with and without the use of exogenous contrast (gadolinium). Cardiovascular MRI (CMR) is the gold standard method for the three-dimensional analysis of cardiothoracic anatomy, the assessment of global and regional myocardial function, and viability imaging (late gadolinium enhancement technique). Using first-pass perfusion imaging under vasodilator stress, CMR has high diagnostic accuracy for the identification of myocardial ischaemia. Oedema imaging using T2-weighted techniques is useful for the identification of acute coronary syndromes and myocardial inflammation. Coronary MR imaging is feasible, and indicated particularly for visualizing anomalous coronaries. Its spatial and temporal resolution is inferior to computed tomography or conventional angiography, and the identification and grading of stenoses remain challenging. Molecular imaging may in future allow visualization of unstable plaque. Novel techniques such as T1- and T2-mapping have recently been developed and offer a quantitative measure of tissue characteristics. CMR also provides important prognostic data for many cardiovascular diseases. CMR is now an essential component of an advanced cardiovascular imaging service, and it is anticipated that its role will continue to grow.

Keywords Cardiac anatomy; cardiac function; coronary angiography; diffuse fibrosis; magnetic resonance imaging (MRI); mapping; oedema; perfusion; viability

Cardiovascular magnetic resonance imaging (CMR) has undergone major technical progress over the last decade. CMR scanning has become faster and more patient friendly, whereas image quality has further improved. A study of cardiac anatomy, (left and right ventricular) function and fibrosis with a modern CMR scanner can be performed within 30 min by an experienced operator. These improvements have led to the widespread adoption of CMR in clinical practice.

Theodoros Karamitsos *MD PhD is Honorary Consultant Cardiologist at the John Radcliffe Hospital, Oxford, UK. Research areas include stress perfusion and oxygenation CMR imaging and the use CMR in the characterization of non-ischaemic cardiomyopathies. Competing interests: none declared.*

Stefan Neubauer *MD FRCP FACC FMedSci is Professor of Cardiovascular Medicine and Honorary Cardiology Consultant at the John Radcliffe Hospital, Oxford, UK. Research areas include most aspects of CMR as well as cardiac metabolism. Competing interests: none declared.*

What's new?

- Cardiovascular magnetic resonance imaging has further advanced in the last four years, becoming faster and more patient friendly
- The evaluation of myocardial oedema with cardiovascular magnetic resonance imaging has undergone further improvement and is now a robust technique applied in everyday clinical practice in many centres
- While still at the research stage, molecular imaging may in future allow specific detection of unstable plaque
- Novel techniques such as T1- and T2-mapping have been developed and provide a quantitative measure of tissue characteristics
- Overall, cardiovascular magnetic resonance has a growing clinical role in ischaemic heart disease and various non-ischaemic cardiomyopathies, with a rapidly expanding pool of prognostic data

Background

Magnetic resonance imaging (MRI) is typically based on the magnetic properties of the hydrogen nucleus, though other nuclei can also be used.[1] In an MRI examination, the patient is placed in a powerful magnetic field with which the protons in the body become aligned. Radio waves in the form of a radiofrequency pulse transmitted into the patient cause the alignment of the protons to change (e.g. by 90°). When this radiofrequency pulse is turned off, the protons in the patient's body return to their neutral position, emitting their own weak radio-wave signals, which are detected by receiver coils and used to produce an image. The phase and amplitude of each returning radio-wave signal can be determined using powerful computers and additional magnetic field gradients, and used to map the position of the excited protons. The resulting image reflects not only proton density, but also the highly complex manner in which protons resonate in their local environment. CMR requires advanced technology, including a high-field magnet (typically 1.5 T, although recently 3.0-T systems are increasingly being used), fast-switching gradient coils, and coils for transmission and signal reception. Compared to other imaging techniques, MRI has a unique ability to perform tissue characterization. Image contrast is influenced by proton density, and T1 and T2 relaxation times, which can vary substantially for different tissues (Box 1). Another way to modify image contrast is by modulating the way the radiofrequency pulses are played out (the MR sequence — Box 2). During an MRI scan, subjects and operators are not exposed to ionizing radiation and there are no known detrimental biological adverse effects of MRI if safety guidelines are followed. The scanner attracts ferromagnetic objects, turning them into projectiles that could lead to significant patient or operator injury and also damage the scanner. The presence of certain medical implants and devices (e.g. pacemakers, defibrillators, cochlear implants, cerebrovascular clips) is a contraindication for routine MR scanning, but nearly all prosthetic cardiac valves, coronary and vascular stents, and orthopaedic implants are safe in a 3-T (or less) MR environment.

Relaxation times in MRI

In MRI, two independent relaxation times are described with respect to the direction of the main magnetic field — longitudinal relaxation and transverse relaxation. Conventionally, these are termed 'T1' and 'T2' relaxation times. Long T1 times reflect slower relaxation parallel to the main magnetic field; long T2 times reflect slower relaxation in the transverse plane. T1 values are typically several times greater than T2 values.

T1 and T2 times vary considerably between different tissues, and these differences are the basis of much of the remarkable contrast resolution of MRI. Tissues with high water content have particularly long T1 and T2 times.

T1 and T2-weighting

- In T1-weighted images, areas with a long T1 time give a low signal. Water-rich areas therefore appear dark.
- In T2-weighted images, areas with a long T2 time give a high signal. Water-rich areas therefore appear bright.

T2* is a time constant describing the exponential decay of signal, due to spin—spin interactions, magnetic field inhomogeneities, and susceptibility effects. T2* is measured by acquiring several T2-weighted images with different echo times. Shorter relaxation times (e.g. from iron loading) cause a more rapid decrease in myocardial SI with increasing echo time, and this rate of decline can be plotted.

Box 1

Claustrophobia may be a problem in a small percentage of patients, and mild sedation usually helps to overcome this. Recently, gadolinium-containing contrast agents have been linked with the development of a rare systemic disorder called nephrogenic systemic fibrosis. The patients at risk for developing this disease are those with acute or chronic severe renal insufficiency (glomerular filtration rate <30 mL/min/1.73 m^2), or acute renal dysfunction of any severity due to the hepatorenal syndrome or in the perioperative liver transplantation period.[2] To date, there is no evidence that other patient groups are at risk. Many MR centres use gadolinium agents that are tightly bound to a cyclic chelate, for which the incidence of nephrogenic systemic fibrosis is near zero. However, it is not known whether immediate haemodialysis protects against nephrogenic systemic fibrosis, so gadolinium-based contrast media should be avoided in high-risk patients unless the diagnostic information sought using contrast-enhanced CMR is essential and not available with non-contrast enhanced CMR or other imaging modalities.

Applications of CMR

Normal and pathological anatomy

Historically, the first application of CMR was the three-dimensional analysis of cardiothoracic anatomy. By providing excellent soft tissue contrast, cardiovascular anatomy can be assessed in virtually any imaging plane (coronal, transverse, sagittal) including individualized double oblique planes. The latter are particularly valuable in complex congenital heart disease. CMR has a very high degree of sensitivity and specificity for detecting diseases of the thoracic aorta such as aneurysm, acute dissection and intramural haemorrhage.[3] It also allows investigation of the consequences of dissection (e.g. thrombosis, aortic incompetence, pericardial effusion) (Figure 1). Thoracic masses found on chest radiography or echocardiography are also

MRI sequences

An MRI sequence comprises a series of radiofrequency pulses that provide the magnetic resonance signal. These are interleaved with a series of field gradient pulses, which provide the spatial encoding of the signal and hence the image.

Spin-echo sequences have traditionally been the 'workhorse' of routine MRI. A 90° pulse is followed by a 180° pulse, and the delay between the two is reflected in the echo time (T_E); the process is repeated after a repetition time (T_R).

- A spin-echo sequence with a short T_R and a short T_E produces an image in which long-T1 areas give a low signal (i.e. black).
- Use of long T_R and long T_E values produces an image in which long-T2 areas give a high signal (i.e. white).
- Long T_R and short T_E values produce a proton density-weighted image.

As a 'rule of thumb', T1-weighted and proton density-weighted images tend to be similar to CT images and are particularly helpful for anatomical orientation. T2-weighted images can be more susceptible to artefacts, but are particularly sensitive to pathological lesions. Many diseased areas appear bright on T2-weighted images, partly as a result of their high water content.

One of the limitations of spin-echo sequences is that they can be relatively slow. Fast spin-echo sequences are similar to conventional spin-echo sequences, but the data are collected faster. Image quality is greater for the same acquisition time, but artefact problems may be worse.

Gradient-echo sequences: in gradient-echo sequences, the 180° pulse is replaced by a reversal of magnetic field gradients. This technique is generally much faster than conventional spin-echo, but can be more prone to artefacts. T1-weighting and T2*-weighting can be undertaken. A refinement of this technique, now widely used to image cardiac function, is termed 'steady state free precession' (SSPF), and provides the highest contrast between chamber blood (white) and myocardium (dark) of all available MR sequences.

Echoplanar imaging: ultrafast techniques, particularly echoplanar imaging, are increasingly being used. They offer short imaging times (e.g. 30—40 ms per slice), but require advanced hardware.

Inversion-recovery sequences: in an inversion-recovery sequence, a 180° pulse is followed by a 90° pulse after an interval TI. An important variant (STIR sequence) uses short TI values to suppress the signal from fat and highlight the signal from many diseased tissues. Another important example is the late enhancement sequence used to image myocardial viability.

Navigator sequences: additional information can be acquired during the image sequence to enable correction for patient movements that would otherwise degrade the image. Use of this group of methods is now moving from the laboratory to newer clinical machines.

Box 2