REVIEW

Specifics of cardiac magnetic resonance imaging in children

Spécificités de l’IRM cardiaque en pédiatrie

Laurent Bonnemains\textsuperscript{a,b,c,d,*}, Francesca Raimondi\textsuperscript{e}, Freddy Odille\textsuperscript{b,c,f}

\textsuperscript{a} Department of Cardiac Surgery, CHU de Strasbourg, 67000 Strasbourg, France
\textsuperscript{b} U947, Inserm, 54000 Nancy, France
\textsuperscript{c} IADI, University of Lorraine, 54000 Nancy, France
\textsuperscript{d} University of Strasbourg, 67000 Strasbourg, France
\textsuperscript{e} Department of Paediatric Cardiology, CHU Necker–Enfants-Malades, 75000 Paris, France
\textsuperscript{f} CIC-IT 1433, Inserm, 54000 Nancy, France

Received 14 September 2015; received in revised form 16 November 2015; accepted 18 November 2015
Available online 14 January 2016

KEYWORDS
MRI; Paediatrics; Heart

Summary This review points out three specific features of cardiac magnetic resonance imaging (MRI) in children: the small size of the heart modifies the usual balance between signal-to-noise ratio and spatial resolution; the higher and more variable heart rate limits tissue characterization and temporal resolution; and motion artefacts (notably respiratory motions) must be dealt with. In the second part of this review, we present the current and future practices of cardiac magnetic resonance (CMR) in children, based on the experience of all French paediatric cardiac MRI centres.

© 2015 Elsevier Masson SAS. All rights reserved.

Abbreviations: 2D, two-dimensional; 3D, three-dimensional; BB, black blood; CMR, cardiac magnetic resonance; CT, computed tomography; ECG, electrocardiogram; GRIC, generalized reconstruction by inversion of coupled systems; LGE, late gadolinium enhancement; MRI, magnetic resonance imaging; Nex, number of excitations; SNR, signal-to-noise ratio; SPECT, single photon emission computed tomography; SSFP, steady-state free precession; TR, repetition time; TSE, turbo spin echo.

* Corresponding author at: Inserm U947, IADI, Tour Drouet, rue du Morvan, 54500 Vandœuvre-lès-Nancy, France.

E-mail address: laurent.bonnemains@inserm.fr (L. Bonnemains).

http://dx.doi.org/10.1016/j.acvd.2015.11.004
1875-2136/© 2015 Elsevier Masson SAS. All rights reserved.
Introduction

Cardiac magnetic resonance (CMR) has developed considerably in the past few years. Hardware and sequences are improving very fast. Higher and more homogeneous fields and stronger gradients allow theoretically higher spatial resolution. Parallel imaging and compressed sensing allow theoretically higher temporal resolution. All conditions seemed aligned to witness the advent of paediatric CMR as one of the most promising available investigation tools in paediatric cardiology [1]. However, paediatric CMR is part of routine clinical practice in only a few centres in Europe. Its own intrinsic limitations and the need for double specific cardiological and radiological expertise make its use difficult in clinical practice. Most cardio-paediatricians are not familiar with the concepts of CMR. The first part of this review aims at clarifying the specific features of CMR in the paediatric population. In the second part, we present the current and future practices of paediatric CMR.

Specifics of paediatric CMR

Three specific features of paediatric CMR will be discussed. Each one is the direct consequence of technical aspects of CMR that encounter limits in children:

- the magnetic resonance signal is produced by the heart, and paediatric hearts are small;
- the magnetic resonance imaging (MRI) process is relatively slow, and paediatric hearts beat rapidly;
- the acquisitions require absence of motion and breath-holding that children cannot comply with.

Heart size and the relationship between voxel size and signal-to-noise ratio

Since paediatric hearts are much smaller than adult hearts, a compromise between two solutions must be chosen. First, the boundaries of the field of view may be reduced to adjust to the child’s anatomy. If the number of voxels of the image is preserved, this will result in a smaller voxel containing fewer protons and ultimately a lower MRI signal. Second, the spatial resolution of the image (size of the k-space matrix) may be reduced. This will result in less informative images. The consequence of the compromise is that the images have a lower signal-to-noise ratio (SNR) and/or a lower spatial resolution. SNR is actually proportional to the product of the voxel dimensions. To counter this, the acquired data can be averaged over multiple excitations, with SNR being proportional to the square root of the number of excitations (Nex) [1]. For instance, let us consider imaging with a single excitation (Nex = 1) and a voxel size set to $1 \times 1 \times 8 \text{mm}^3$; if the voxel size is reduced to $0.7 \times 0.7 \times 8 \text{mm}^3$, the same SNR will be obtained with Nex = 4, which implies increasing the scan time fourfold.

Another element that is of utmost importance with regard to SNR is the receiver coil. MRI scanners are equipped with a great variety of multiple-channel surface coil arrays to fit all shapes of the average adult anatomy (head, torso, knee coils, etc.). However, they cannot be adapted to the dimensions of each individual patient. Radio-frequency coil receivers can be thought of as simple coil loops. The diameter of these loops should be large enough to capture signals from protons deep inside the body, but as small as possible to capture less noise coming from the rest of the body. In general practice, only conventional adult coils are available, leading to a suboptimal SNR. Dedicated or scalable coils would be an interesting field for future research.

Heart rate and cardiac synchronization

The normal heart rate in infants (90–180 bpm) is higher than in adults (60–100 bpm). For cardiac MRI, higher heart rates have two general consequences. Firstly, the heart rest phase (diastasis in mid-diastole) shortens and disappears after 90 bpm [2]. Shorter diastasis implies that a smaller portion of the k-space can be acquired during each cardiac cycle to avoid motion blurring. The use of end-systole (40% of the cardiac cycle), instead of mid-diastole (75% of the cardiac cycle), has been advocated when heart rate is >70 bpm [3], but this has not been validated in children. The use of systole may not be compatible with all preparation pulses as they may require a certain amount of time before the readout. Those pulses can be performed in anticipation, during the previous cardiac cycle, but it could require a prospective guess of the next cardiac cycle length [4]. Secondly, the cardiac cycle length becomes very short with regard to the corresponding cardiac time constants (T1 and T2). For T1-sensitive sequences, it is preferable to wait between consecutive MRI excitations (ideally $3 \times T1$ of the organ of interest, i.e. at least 2–3 s or several
heartbeats) to leave enough time for the magnetization of the protons to come back to its equilibrium state. Another specific issue of the paediatric heart rate is its variability on a beat-to-beat basis. Reconstruction of images acquired during cardiac cycles of variable length is more complex. Temporal resolution is affected by this reconstruction and velocity measurements may be altered [5].

### Motion artefacts

Children’s respiratory motion and small duration of mid-diastole diastasis cause important blurring and motion artefacts that impair the reliability of tissue characterization sequences, such as T1 or T2 mapping or fibrosis detection. Therefore, many paediatric teams do not use CMR for tissue characterization or resort to deep sedation or general anaesthesia [6], although this entails its own risks and requires an MRI team trained in sedation, airway management and caring for cardiac patients. When sedation is not performed, alternative solutions are useful to obtain the child’s cooperation, such as playing music or showing films (Fig. 1) [7]. Several solutions have been proposed to cope with respiratory motion. The simple averaging of several MRI datasets is possible when the breathing is calm and periodic (notably for infants) [1]. For older children, the results are rather uncertain and it is often necessary to perform several acquisitions before obtaining clinically relevant information. Respiratory gating is feasible but is also inefficient because it increases the acquisition duration whereas the acquisition timing is often constrained, notably by the kinematics of the gadolinium contrast agent within tissues. Recently, very fast acquisitions have been proposed with low SNR. Several teams have proposed motion-corrected reconstructions [7].

![Figure 1](image-url)  
Example of a solution to obtain the cooperation of children: oblique glasses (plain arrow) and video projection on a wood-frame screen (dashed arrow) [7]. Reproduced with permission from BioMed Central.

### Current practice of paediatric CMR

Cardiac MRI in children with congenital heart diseases may add important elements to echocardiographic data in terms of anatomy, haemodynamics and tissue characterization (Table 1).

### Anatomy

Two sequences — three-dimensional (3D) steady-state free precession (SSFP) and contrast-enhanced angiography — are widely used to study cardiac anatomy. By coupling these two sequences, invasive and/or irradiating examinations (e.g. cardiac catheterization or cardiac computed tomography

| Table 1 | Indications for paediatric CMR.          |  |
|---------|------------------------------------------|  |
| Objective | Best solution | Alternative solution |
| **Anatomy** |  |  |
| Extracardiac |  |  |
| Age < 5 years | Gadolinium 3D angiography | T2 TSE BB |
| Age > 5 years | 3D SSFP | T2 TSE BB, gadolinium 3D angiography if stent |
| Intracardiac |  |  |
| Age < 5 years | 2D SSFP |  |
| Age > 5 years | 3D SSFP |  |
| **Quantification** |  |  |
| Flow | Phase-contrast (free-breathing if necessary) |  |
| Volumes/EF | 2D SSFP (free-breathing if necessary) |  |
| **Tissue characterization** |  |  |
| Oedema |  |  |
| HR > 110 bpm | 2D SSFP | T2 TSE |
| HR < 110 bpm | T2 mapping | T2 TSE |
| Fibrosis |  |  |
| HR > 110 bpm | Gadolinium + 2D SSFP | T1 TSE |
| HR < 110 bpm | T1 mapping | LGE, T1 TSE |

2D: two-dimensional; 3D: three-dimensional; BB: black blood; CMR: cardiac magnetic resonance; EF: ejection fraction; HR: heart rate; LGE: late gadolinium enhancement; SSFP: steady-state free precession; TSE: turbo spin echo (or fast spin echo).
Contrast-enhanced angiography (CT) angiography can often be avoided for morphological studies.

3D SSFP

3D SSFP is an electrocardiogram (ECG)-triggered pulse sequence with respiratory motion compensation by diaphragmatic navigators or navigation on the heart itself, called self-navigation. It is acquired in free-breathing. This type of sequence may be used with or without contrast medium to produce high-resolution 3D data of the whole heart and extracardiac structures. However, to obtain good image resolution, regular respiration and regular heart rate are mandatory. Therefore, its use is difficult in young children and neonates. However, this sequence is very powerful for defining congenital heart disease, to precisely visualize the heart segmentation and to classify the cardiopathy properly. The acquired full volume is isotropic and allows reconstruction of the heart following any oblique plane, to better identify the correct surgical strategy in complex congenital heart diseases. Part of the thoracic anatomy can be isolated and presented as a 3D object, such as the post-surgery aorta coarctation presented in Fig. 2A. The heart can also easily be 3D-printed such as in Fig. 2B. It is also helpful to define the relationship between the heart, the great vessels and thoracic structures. For children, structures such as ostia of coronary arteries and their proximal segments (e.g. in Fig. 2C and D) or aorto-pulmonary collaterals are well visualized. However, the spatial resolution is not sufficient to visualize the details of mitral or tricuspid valve anatomy. Its other pitfall concerns visualization of structures containing a stent that causes important artefacts on its proximal structures.

Contrast-enhanced angiography

Contrast-enhanced angiography is complementary to 3D SSFP, especially in patients with high heart rate and irregular breathing. It needs intravenous gadolinium-based contrast agents to reduce the T1 relaxation time of blood. The acquisition is made on 1–3 different times of the cardiac cycle (1–2 if acquired in free-breathing) to visualize the different phases of contrast filling: superior vena cava, pulmonary arteries, aorta (e.g. in Fig. 3A), pulmonary veins and inferior vena cava. The time of acquisition is important according to different indications to visualize different anatomical structures. Stents and small vessels, as well as distal pulmonary arteries and aorto-pulmonary collaterals, are well visualized, as presented in Fig. 3B.

Haemodynamics

In the majority of cases, morphological studies need to be completed by haemodynamic data.

Blood flow

With the velocity-encoded phase-contrast cine sequence, it is possible to measure the volume of blood flow passing through any vessel of interest by placing a slice perpendicular to it. With specific software, it is possible to calculate the volume and the direction of the blood stream to estimate the forward and backward volume and to calculate valve regurgitation. It is important to repeat this measurement within

![Figure 2](image)

Examples of current usage of 3D SSFP sequences for anatomy in children. A. A 3D reconstruction of the aorta of a 6-year-old boy several years after surgery for coarctation, based on a 3D SSFP acquisition without any gadolinium injection. B. A partial 3D printing of a complex double outlet right ventricle in a child of weight 5 kg. The tunnel from the left ventricle to the aorta has been outlined in white. C and D. 3D SSFP images of the coronary arteries in a 5-year-old child with transposition of the great arteries. The right coronary (C) is normal whereas the left coronary ostium (D) has been enlarged by a patch in a second surgery 1 month after the arterial switch. 3D: three-dimensional; Ao: aorta; LV: left ventricle; SSFP: steady-state free precession; RA: right atrium; RV: right ventricle; VSD: ventricular septal defect.

![Figure 3](image)

Examples of current usages of gadolinium-enhanced angiography for anatomy in children. A. A 3D reconstruction of the aorta of a 6-year-old child with coarctation, based on a gadolinium-enhanced 3D angiography. B. An axial view (based on a 3D angiography) of the pulmonary trunk and proximal pulmonary branches in a 12-year-old child with a prosthetic pulmonary trunk dilated by a stent (white arrow). A second, smaller, stent has been positioned at the entrance of the left branch. 3D: three-dimensional.
all of the principal veins and great vessels, and also with a phantom to correct errors and to test data consistency [8]. This sequence is reliable for large vessels, therefore pulmonary and aortic flow and regurgitation are fairly reliable. However, it is less reliable for small vessels such as pulmonary veins or little collateral vessels, and rather difficult to use for moving structures such as atrioventricular valves [9,10]. This sequence can easily be used with free-breathing children or neonates with a simple averaging. It may be used to evaluate shunts (intracardiac or extracardiac, except arteriovenous shunt) and regurgitation [11].

Ventricular function and volumes
Ventricular function and volumes are estimated by two-dimensional (2D) SSFP cine sequence. This sequence may be acquired on any desired plane to study the global and regional functions in specific cardiopathies and in cardiomyopathies of the right and/or left ventricle. Volumes are calculated on short-axis views according to the international consensus policy [12]. This sequence is used to assess regional function of the right ventricle better than echocardiography, so it is recommended in the study of pathologies affecting the right heart (cardiomyopathies or post-surgical congenital heart disease like Tetralogy of Fallot). It is acquired normally in breath-holding but it is possible to obtain good quality images in free-breathing small children with a simple averaging (multiple Nex).

Myocardial perfusion
Myocardial perfusion at rest and/or after pharmacological stress (dipyridamole or adenosine) may be evaluated by a first-pass perfusion imaging technique after gadolinium-based contrast medium injection. Stress perfusion CMR has been extensively validated in adults affected by ischaemic heart disease. It is superior to single photon emission computed tomography (SPECT) and it has comparable diagnostic accuracy to positron emission tomography and fractional flow reserve [13]. Moreover, compared with SPECT, CMR provides information about coronary artery anatomy, so it is a valid option to assess myocardial ischaemia. Its use is restricted to children in our experience.

Tissue characterization
The principal peculiarity of CMR in comparison to other imaging techniques is the possibility to perform tissue characterization. Three cardiac components are usually detected: fat, oedema and fibrosis. Generally, fat has low T1 and high T2, fibrosis has elevated T1, and oedema has high T1 and T2. To increase the contrast between fibrosis/oedema and normal myocardium, the use of a contrast agent (e.g. gadolinium) is recommended [14]. Tissue characterization is very difficult to achieve in neonates, and it should be restricted, according to our experience, to infants and children with heart rates < 110 bpm (for quantification or late gadolinium enhancement [LGE]) and with breath-hold capacities (for T1 or T2 quantification). Two types of analysis can be performed:

- quantification of T1 and T2 can be performed by T1 and T2 mapping sequences. These sequences are currently available on most MRI scanners. The images are usually coloured and each voxel colour directly codes for a T1 or T2 value (e.g. in Fig. 4A). Each tissue (notably fat, oedema and fibrosis) can be recognized by analyzing the same zone on T1 and T2 maps. T1 mapping sequence soon after gadolinium injection is technically challenging and several artefacts often make the interpretation quite difficult. In our experience, accurate quantification requires breath-holding capacities and heart rate < 110 bpm;

- T1- and T2-weighted images are specific sequences designed to visually discriminate tissues according to their T1 or T2. T1-weighted images have low repetition time (TR) (10–20 ms) so that tissues with short T1 appear brighter. T2-weighted images have long echo times (TE and TR > 40–50 ms) so that tissues with long T2 appear brighter. LGE sequences are T1-weighted images with a specific choice of the delay between the inversion pulse and the readout to ensure that normal myocardium appears dark. In our experience, T1- or T2-weighted images can be acquired even in young children (breath-hold is not mandatory) but LGE requires heart rate < 110 bpm. LGE is very informative when large zones of the myocardium are infarcted, but its interpretation in children with diffuse fibrosis may be complex (e.g. in Fig. 4B [7]). 2D SSFP constitutes a specific sequence with
T2/T1-weighting that is very robust in children. In clinical practice, SSFP cine sequences just before and soon after gadolinium injection is a good alternative to detect oedema/fibrosis, as illustrated in Fig. 4C [7]. Oedema resulting in longer T2 results in spontaneously hyper-intense tissue before injection. This spontaneous signal must be compared with the enhancement after contrast injection, which is due to the gadolinium reduction in T1. In theory, it is possible to discriminate the oedema and fibrosis zones, but in practice, this distinction is complex and LGE at an early stage often decreases with time [15].

Tissue characterization has a high potential added value for the diagnosis and characterization of acquired or congenital cardiomyopathies, inflammatory processes (myocarditis, graft rejection [16], etc.) and cardiac tumours.

**Future of paediatric CMR**

**Novel accelerated imaging techniques**

Beyond commercially available acceleration techniques such as partial [17] and parallel [18] acquisition of the k-space, novel recent developments are expected to benefit paediatric CMR. In particular, the recent theory of compressed sensing, applied to MRI, states that an image can be reconstructed from a reduced subset of randomly chosen k-space samples, which means the acquisition can be further accelerated. The main condition to be able to apply compressed sensing is that the image should be compressible (e.g. the background of the image does not contain any useful information). This technique has been applied to body (chest and abdomen) paediatric MRI [19].

**Advanced motion correction**

Recent motion correction techniques allow retrospective correction of complex organ motion, including non-rigid or elastic deformations. Generalized reconstruction by inversion of coupled systems (GRICS) [20,21] is one such algorithm, able to produce clinically relevant images from MRI data corrupted by respiratory motion with the help of respiratory motion sensors (two respiratory pneumatic belts positioned on the thorax and the abdomen) connected to a recording system. Acquisitions are performed in complete free-breathing and images are reconstructed retrospectively with a non-rigid motion correction based on all data (no respiratory gating) and adapted to the patient’s specific motion. Preliminary results of the application of GRICS to CMR of children and young adults have been shown in Duchenne muscular dystrophy patients [7]. Closely related approaches have been shown in paediatric abdominal applications that combine advanced motion correction with compressed sensing [22].

**Foetal heart imaging**

Ultrasound is the recommended technique as the primary approach for the assessment of foetal cardiac structures. Factors that limit ultrasound, such as maternal obesity, foetal position and oligohydramnios, do not necessarily compromise the image quality of MRI. Therefore, cardiac MRI has been investigated in various centres, but remains mainly restricted to research. Foetal cardiac MRI constitutes a complex challenge due to rapid foetal heart rate, small heart size, foetal movements and because real ECG-gating of the foetal heart is not possible. Indeed, although the foetal ECG is visible on the mother’s ECG [23], it is typically an order of magnitude smaller in amplitude and might not be large enough compared to the interferences caused by the electromagnetic MRI environment (magneto-hydrodynamic effects caused by blood flows perpendicular to the static magnetic field and gradient-switching artefacts during the sequences). Therefore, QRS detection is very difficult on the foetal ECG in MRI. Different solutions have been proposed for ECG triggering with varying results: no triggering [24], external ultrasound captor [25] and metric optimized gating with retrospective reconstruction [26]. These solutions are not yet available in clinical practice.

**Interventional MRI**

Cardiac catheterization entirely guided by MRI was reported in 2003 [27]. It requires specific non-magnetic equipment during the procedure. Several aspects of catheterization are much more complex (plastic catheters are difficult to see in MRI, metallic guide-wires are prohibited, etc.). However, MRI allows a simple evaluation of flows and simplifies the computation of pulmonary resistance. The higher cost of such dedicated environments constitutes a real drawback and it is not yet clear whether this technique will become more widespread.

MRI may also be used to monitor temperature during radio-frequency ablation, but this is rarely used in paediatric populations. MRI could also be a way to deliver energy locally by means of focalized radio-frequency.

**Conclusions**

Paediatric cardiac MRI is a challenging modality with tremendous potential. It is non-invasive and may be repeated several times without harm to the patient. It provides anatomical, haemodynamic and pathological information and is rapidly acquiring a key role in the diagnostic process of congenital heart diseases.

**Disclosure of interest**

The authors declare that they have no competing interest.

**References**


