

CONGENITAL HEART DISEASE

MR Evaluation of Cardiovascular Physiology in Congenital Heart Disease: Flow and Function

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ABSTRACT

Cardiovascular magnetic resonance (CMR) has become the method of choice in the evaluation of a number of questions in congenital heart disease. In addition to morphology, modern CMR techniques allow the visualization of function and flow in a temporally resolved manner. Among the pathologies where these methods play a major role are shunts, septal defects, aortic coarctation, anomalies of the pulmonary arteries, and valvular regurgitation. This paper explains the basics of functional and flow encoded CMR and discusses their application in the assessment of several types of congenital heart disease.

INTRODUCTION

Cardiovascular magnetic resonance (CMR) has undergone considerable development during recent years. Improvements in hardware and development of new acquisition techniques have strengthened the modality to become the method of choice in many cardiovascular questions. Pediatric CMR profits from many of those improvements as well, allowing for better MR images in shorter acquisition times and leading to comprehensive evaluation of cardiac morphology and function in a short time.

Evaluation of the morphology of congenital heart diseases was one of the first major applications of CMR. The absence of ionizing radiation and of iodinated contrast agents is particularly beneficial in pediatric cases and thus encourages substituting it for x-ray imaging. Early applications of CMR in children focused on patients in which echocardiography was not able to provide a reliable diagnosis. However, faster MRI and more

Keywords: Congenital Heart Disease, Phase Contrast MRI, Cine MRI, Myocardial Tagging, Review. Received 2 August 2005; accepted 1 April 2006. Correspondence to: Oliver Weber, PhD University Hospital and University of Basel Department of Medical Radiology Petersgraben 4 CH-4031 Basel Switzerland phone +41 585 563 216 fax +41 585 542 363 email: oweber@gmail.com widespread availability of imagers capable of CMR have made it more competitive with echocardiography in all aspects of congenital heart disease.

High-resolution morphological information is typically obtained from thin-slice, T1-weighted, black-blood images and contrast-enhanced, three-dimensional (3D) MR angiography. Additionally, functional information provided by, and in many cases unique to, MRI has been proven to be of major value. Information on blood flow in the major vessels such as aorta, pulmonary arteries, central veins, and surgical conducts, is obtained from velocity-encoded cine (VEC) MRI. Information on tissue motion is readily available from cine steady-state freeprecession (SSFP) MRI in arbitrary orientations. Finally, for a more accurate evaluation of tissue motion and calculation of strain data, myocardial tagging is available.

In this review, the techniques for functional CMR are discussed, and some clinical applications are presented.

TECHNIQUES

Segmentation and synchronization to the cardiac cycle

Information on function and flow is obtained from timeresolved series of gradient-echo images, each time frame corresponding to a phase of the cardiac cycle. Since these images are frequently viewed as a movie, these techniques are also known as "cine MRI" techniques. In most cases, the spatial and temporal resolution required makes it impossible to acquire a full series in a single heart beat. Instead, each image is assembled from data acquired during several heart beats (segmentation). In order to attribute the data to the proper cardiac phase, an electrocardiogram (ECG) or peripheral pulse must be recorded during the MRI acquisition. After completion of data acquisition, data are re-ordered to allow for reconstruction of a complete image of each cardiac phase. Theoretically, the time interval between two cardiac phases can be chosen as short as a single repetition time (TR), i.e., in the order of 5 ms. However, in order to reduce acquisition time and data size and because a temporal resolution in the order of 50–75 ms per phase has been found to be sufficient for most clinical applications, the cardiac cycle is typically divided into 16 phases. Only at very low heart rates must a higher number of phases (e.g., 24 or 32) be acquired for appropriate temporal coverage (1).

For synchronization to the cardiac cycle, an ECG or a peripheral pulse unit (PPU) is used. Generally, the R-wave of the ECG allows more reliable triggering and thus results in more consistent phases over the course of the entire imaging time. Furthermore, it demarcates the beginning of the systole and thus serves as a well-defined and physiologically meaningful starting point for imaging. In the presence of a strong magnetic field, the ECG trace may be severely distorted by the magnetohydrodynamic effect. Radio-frequency pulses and the switching magnetic gradients also interfere with the ECG. In the presence of these artifacts, the four-lead vector ECG provides more reliable detection of the R-wave and thus better cardiac synchronization than the three-lead (scalar) ECG (2). The PPU signal suffers from more variability than the ECG and may lead to wrongly attributed data, resulting in image blurring and less reliable results. PPU is therefore only recommended in patients in whom a reliable ECG-trace is not available.

Synchronization to the cardiac rhythm is performed either prospectively or retrospectively. In retrospective gating, MR imaging is performed continuously. At the end of each cardiac cycle, the acquired image data is attributed to their respective phases, and the next part of the data is collected. Main advantages of retrospective gating are preservation of steady state (which is especially important in SSFP sequences) and coverage of the entire cardiac cycle (without missing data at the beginning and at the end of the cardiac cycle). Therefore, retrospective gating is preferred for cine imaging of cardiac function. In prospective triggering, the system waits for the trigger impulse before acquiring data for the cardiac phases. This method allows inclusion of shared prepulses, such as spatially selective saturation pulses for myocardial tagging. However, due to the waiting period at the end of the cardiac cycle, no data of the time point immediately before the trigger event is available.

For all cine MRI sequences, reliable synchronization and a steady heart rate are important. Large variations in R-R intervals (either true ones, caused by arrhythmias, or erroneous ones caused by missed or wrongly detected trigger events) can cause data from different heart phases to be combined in a single image, resulting in artifacts and image degradation. Mechanisms to detect arrhythmias and reject data acquired during those heart beats may improve image quality, at the cost of prolonged scanning times. To prevent image artifacts caused by respiratory motion, many cine sequences are performed during a breathhold. Alternatively, signal averaging can be used to reduce the severity of the artifacts.

In patients with severely irregular heart rhythm, segmented cine MRI is not possible. In these patients, it may be necessary to perform a real-time sequence without cardiac synchronization. In real-time sequences, all images are acquired in a single-shot mode, i.e., all data for each image is acquired in a single burst. In comparison with segmented sequences, typically both spatial and temporal resolution are reduced, but in many cases images are of sufficient quality to give an impression of cardiac function. Real-time sequences may also be advantageous in infants and young children who can not hold their breath.

Imaging of basic function

In recent years, steady-state free-precession (SSFP) sequences (also known as balanced FFE, TrueFisp, and FIESTA) have practically replaced previously used spoiled gradient echo sequences for the imaging of cardiac function. The properties of SSFP make it extremely useful for imaging of the cardiac function. Strong T2/T1-weighted and homogeneous contrast shows cavitary blood bright and myocardium with lower signal intensity (Fig. 1). Contrast is virtually independent of inflow, allowing imaging at consistent quality in all orientations, including short, horizontal, and vertical long axes, and in orientation of the outflow tracts. Since signal and contrast are independent of echo time (TE) and repetition time (TR), TR can be shortened as much as gradient switching times, gradient strength, and limitations on radiofrequency absorption rate allow. SSFP sequences are therefore also fast, allowing acquisition of a single slice with 16 heart phases in a breathhold of a few seconds duration.

For ventricular volumetrics, a number of adjacent images are acquired in the short axis orientation covering the entire heart from apex to base (Fig. 1). For each short axis slice, a region of interest (ROI) is defined by outlining the endo-cardial contour of the left ventricle (LV) or the right ventricle (RV) both on the end-systolic and end-diastolic images. For each slice the area of the ROI is multiplied by the slice thickness. Summation over all slices results in end-systolic volume (ESV) and end-diastolic volume (EDV), respectively. The difference between the two volumes corresponds to the stroke volume (SV). From these numbers, ejection fraction is calculated as EF = SV/EDV. Cardiac index is calculated as cardiac output per minute, normalized by the body surface area, i.e., $CI = SV^*HR/BSA$, where HR is the heart rate in beats per minute.

To measure ventricular mass, the epi-cardial ROI is traced on each slice to define an area that includes both myocardium and ventricular cavity. A second ROI is drawn along the endo-cardial border, defining the area of the ventricular cavity. Subtraction of the latter from the former yields the area of myocardium on each image. The area of myocardium is multiplied by the slice thickness and summed over all slices to give the myocardial volume. To calculate myocardial mass, the myocardial volume must be multiplied by the density of myocardium (1.05 g/mL).

Once the myocardial wall is defined, also radial wall thickness can be calculated. If this is done on both end-systolic and





end-diastolic images, the change in wall thickness indicates wall thickening.

Several studies have shown very small values for interstudy variability of MRI for quantifying ventricular volume and mass (3–5). This renders it the optimal technique for monitoring ventricular function in patients over time and offers a distinct advantage to echocardiography, especially when evaluating RV volume and function.

Cine MR images are also employed to evaluate valvular function (6). A plane approximately parallel to the valve leaflet or cusps can demonstrate valve motion, such as the horizontal long axis plane for assessing mitral valve motion. The high velocity jet caused by valvular stenosis and regurgitation may be identified on SSFP images as flow void (dark structures caused by intravoxel phase dispersion in the presence of turbulent flow; Fig. 2). Even though no volumetric numbers of jets are available with this method, extent and duration of the jet allow drawing semi-quantitative conclusions on the severity of the defect (7).

SSFP sequences work best with short TR; fast switching and strong magnetic gradients are thus preferable. With longer TR's,

off-resonance artifacts (banding artifacts) may distort the image. For the same reason, appropriate shimming is required. Volume shimming in the volume of interest (e.g., the heart) is therefore recommended. Other potential artifacts include signal void due to turbulent flow; artifacts caused by moving spins (8); and outof-slice effects (9).

Velocity-encoded MRI

Quantification of blood flow is done using velocity-encoded cine MRI (VEC-MRI). Since this method is based on the principle of phase shift, it is also called phase-contrast imaging. Magnetic spins moving along a magnetic gradient at a constant velocity accumulate a phase shift proportional to the velocity. In the phase map, which is reconstructed in addition to the (normally reconstructed) modulus image, the signal intensity (gray value) signifies the average velocity of each pixel. In order to show velocity patterns throughout the cardiac cycle, an ECG-gated cine gradient-echo sequence is employed with flow-encoding gradients inserted along the desired direction. Each of the



Figure 2. A single phase of a dynamic four-chamber view (acquired using an SSFP sequence) demonstrates a jet from the left ventricle into the left atrium, indicating mitral valve insufficiency.

resulting phase maps represents velocity at a different time point in the cardiac cycle. Flow-encoding in arbitrary directions is theoretically possible. In practice however, options are typically restricted to the three orthogonal main axes: perpendicular to the imaging plane (i.e., in slice-selection direction), as well as parallel to the imaging plane (i.e. in measurement and phaseencoding directions). Encoding perpendicular to the imaging plane results in information on through-plane velocity. This is used to determine volumetric flow data by imaging the vessel of interest in cross-section. From this set of images, instantaneous flow (i.e., mL/sec during the respective heart phase) is calculated by drawing an ROI around the vessel and integrating velocity over the vessel area in each time frame. A flow curve is obtained by plotting these data against time (Fig. 3). Integrating flow values over the entire cardiac cycle (which is equivalent to calculating the area under the curve) results in stroke volume; subsequent multiplication by the heart rate gives flow rates in mL/min. This can, for instance, be used to calculate the cardiac output by assessing flow through the aortic valve. For volumetric flow, the imaging plane should be chosen perpendicular to the vessel of interest. However, the errors made in through-plane velocity and area compensate each other in case the imaging plane is defined at an angle slightly off a perfect perpendicular orientation (10), resulting in proper volumetric data. The peak velocity however will be underestimated in this case.

Encoding of in-plane motion, with the vessel imaged in plane, allows visualization of flow patterns (Fig. 4) and assessment of peak velocity in a region. This information can be used to calculate pressure gradients using the modified Bernoulli equation (peak pressure gradient = $4 \times$ peak velocity²). Potential sources of underestimation include misalignment (the flow not being accurately along the flow encoding direction), partial volume effects (averaging over different velocity components within the same pixel), and signal loss due to turbulent flow. Despite these potential problems, in-plane flow imaging provides appropriate



Figure 3. A velocity-encoded gradient echo sequence was used to image ascending and descending aorta in a normal volunteer. The magnitude image (left-top) shows the ascending and the descending aorta in cross-sectional views. The main pulmonary artery winds around the ascending aorta and bifurcates into the right PA and (out of plane) the left PA. The phase image (velocity map) shows through-plane flow velocities approximately 160 ms after the R-wave, during maximum flow in the ascending aorta. The blood in the ascending aorta shows positive velocity (equivalent to flow in foot-head direction; bright signal), whereas the blood in the descending aorta shows negative velocity (equivalent to flow in head-foot direction; dark signal). Non-moving structures, such as the chest wall, as well as pixels without sufficient signal contributions, show medium-value gray (equivalent to zero velocity). venc was set to 200 cm/s.

Flow values (in mL/s) are calculated by multiplying mean velocity in the vessel by the vessel area. The graph shows flow values over all cardiac phases. Note the greater net flow (=area under the curve; 6.7 L/min), as well as greater peak flow, in the ascending aorta as compared to the descending aorta (net flow, 4.1 L/min).



Figure 4. Magnitude view showing the aortic arch (A); in-plane velocity-encoding (venc = 200 cm/s) in readout-direction (foothead; B) and in phase-encoding direction (anterior-posterior; C); and through-plane velocity-encoding (venc = 50 cm/s) in slice-selection direction (left-right; D). From these velocity values, 3D-velocity patterns can be calculated according to the rule of Pythagoras. Note the phase aliasing in the pulmonary artery in the through-plane flow-encoded image due to the smaller venc.

The phase-images with in-plane velocity-encoding demonstrate flow towards the head in the ascending aorta (bright signal in B), towards the back in the arch (bright signal in C), and towards the feet in the descending aorta (dark signal in B). Image D illustrates the helical flow in the aorta.

visualization of flow patterns and generally gives a relatively accurate estimate of peak velocity.

Acquisition of time-resolved 3D flow data in a 3D-volume can be used to investigate complex flow patterns in major vessels such as the aortic arch (11) or distal to aortic valve prostheses (12). In normal volunteers, a right handed helical blood flow is normally seen in the aortic arch. In patients with aortic aneurysms, coarctation, and other pathologies, more complex and potentially pathologic flow patterns might be seen, such as complex circular flow in aneurysms (11). Downstream of aortic valve prostheses, distinct flow patterns were observed that reflected the valve design. During early flow acceleration, reverse flow occurred adjacent to high velocity jets. Later in systole, flow became confined to the central vessel area and reverse flow along the inner aortic curvature developed (12).

Velocity encoding reflects velocity as phase shift in the phase images. Since only phase shifts in the range -180° to $+180^{\circ}$ can be distinguished, only velocities within a certain range can be properly distinguished. This range is defined by setting the encoding velocity (v_{enc}) during scan definition. Velocities outside this range will be shown with a phase shift identical to a velocity in the range (and therefore offset by a multiple of 2 v_{enc}). This artifact called phase-wrap is usually readily discernible due to its sharp border between wrapped and non-wrapped pixels in the phase image (Fig. 5). Choosing the v_{enc} considerably higher than the maximum velocity present in the vessel of interest on the other side results in exploitation of only part of the encoding range and in reduced accuracy. For optimal results, the v_{enc} should therefore be chosen slightly higher than the maximal velocity expected in the vessel of interest.

All velocity measurements using the phase contrast mechanisms are subject to errors caused by concomitant gradients (13), resulting in higher-order effects on the phase (so-called Maxwell terms). These terms grow with increasing gradient strength; therefore magnetic gradients should be kept small for the assessment of flow. However, small gradient amplitudes result in long echo times and therefore in signal loss in turbulent or fast flow, making impossible the detection of accurate phase values in these regions. In this case, switching to higher gradients shortens echo time and thus reduces artifacts caused by phase dispersion. However, it must be kept in mind that the measurements are less accurate than measurements obtained with lower gradients.

Myocardial tagging

Myocardial tagging is a method for more thorough investigation of cardiac motion that is mostly used in research applications. It is able to provide detailed information on translation, rotation, and deformation and allows calculation of such measures as wall motion, regional wall thickening, and strain. This is achieved by destroying coherent magnetization periodically over space and subsequently performing a cine acquisition using a gradient echo sequence. Over the cardiac cycle, the saturated tissue does not give any MR-detectable signal and thus shows up black on the MR image. By applying a periodic pattern of saturation stripes onto the tissue, motion can accurately be followed during cardiac contraction and dilatation (Fig. 6). This can be done in a single dimension (line-tagging) with subsequent multiplication of two data sets with orthogonal tagging direction to create a grid-pattern, or directly in two dimensions (grid tagging). Common procedures for application of the tagging pattern include spatial modulation of magnetization (SPAMM; [14]) and complementary SPAMM (C-SPAMM; [15]). For SPAMM, a series of saturation pulses applied at the beginning of the cardiac cycle destroys magnetization in periodic planes perpendicular



Figure 5. Venc-MRI (venc = 100 cm/s) acquired in axial orientation, showing ascending and descending aorta in cross-sections. The phase images acquired during systole 160 ms (left) and 200 ms (right) after the R-wave demonstrate phase-wrapping (arrows) due to blood flowing faster than the venc-value (103 cm/s in the ascending aorta; -116 cm/s in the descending aorta). Typically phase-wrapping is easily recognized on the velocity maps due to sharp edges between wrapped and non-wrapped pixels. If phase-wrapping occurs, the acquisition should be repeated with higher venc.



Figure 6. Myocardial tagging magnetically marks the tissue, resulting in black lines in the MR image. When imaged with a cine sequence, the dislocation of the tag lines can be followed, visualizing tissue motion. The MR images show C-SPAMM short axis images during end-systole (left) and end-diastole (right), illustrating both contraction and rotation. Harmonic phase (HARP) processing allows for extraction of a number of quantitative measures, such as circumferential shortening and rotation, in different regions of the myocardium. Tagging data courtesy of S. Ryf, ETH Zurich and University of Zurich, Switzerland.

to the imaging slice, resulting in parallel black lines on the subsequently acquired images. C-SPAMM consists of two separate scans (performed in a single breathhold) with complementary magnetization, but identical regions of signal saturation. Subsequent subtraction preserves the black lines, but increases the signal obtained in the remaining tissue. C-SPAMM requires longer acquisition duration and thus longer breathhold duration, but provides longer tag-persistence and thus also allows investigation of early- to mid-diastole. In both methods, the tag lines decay over time (T1-effects) and with the application of RF pulses (for imaging excitation). Low flip angles and reduced number of excitation pulses (such as available with echo planar imaging [EPI] or spiral imaging [16]) may thus be preferable.

Evaluation of the tagging images requires detection of the tag lines in some way. This can be done on the MR images by defining the lines or intersections of lines manually or semiautomatically. However, these procedures are cumbersome and prone to inaccuracies and poor reproducibility. Recently an automated and reliable method was presented called HARP (harmonic phase), which exploits the phase shift in the MR images caused by the periodic saturation pattern (17). Only minimal user interaction is required, resulting in reproducible results. From the tracked grid-intersections, wall motion and its rotational and radial components, as well as wall thickening and wall thinning can be deduced.

In addition to following the motion of tissue over time, tagging allows strain analysis based on finite element methods. Typically, the tracked points are triangulated and subsequently, e.g., radial and tangential strain components are calculated. Just like motion parameters, strain parameters can be mapped onto an image to generate an intuitive display of the numbers. Among other applications, myocardial tagging has been used to investigate function in patients with single left ventricle (Fig. 7) (18) and transplanted hearts (19).



Figure 7. Early diastolic (left) and late diastolic (right) SPAMMtagged short axis images of a patient with a single ventricle. From the tagging images, motion parameters such as twist and radial motion can be determined, which subsequently allow for calculation of (circumferential and radial) strain. As compared to the normal left ventricle, twisting motion was found to be reduced in single ventricle. Furthermore, the septal wall showed reduced circumferential shortening strain and increased radial strain. Note that for the images shown the tagging pattern was applied during early diastole. Images courtesy of M.A. Fogel, The Children's Hospital of Philadelphia, PA.

CLINICAL APPLICATIONS

Left-to-right shunts

Functional and VEC-MRI are useful for the evaluation of left-to-right shunts, especially for partial anomalous pulmonary venous connection, sinus venosus, atrial septal defects (ASD) and supracristal ventricular septal defects (VSD). These lesions may be more problematic for echocardiography compared to other left-to-right shunts. Functional SSFP cine imaging in many cases shows a systolic flow jet into the right heart. It also allows for quantification and comparison of ventricular stroke volumes of the left and right sides. VEC cine MRI may be employed to measure the pulmonary and aortic flow and to calculate pulmonary to systemic flow ratio (Qp/Qs).

Shunts are quantified by means of comparing rightventricular and left-ventricular stroke volumes, either determined from functional cine MRI data or from VEC-MRI data.

Atrial septal defect

Spin echo and cine MR images in the transaxial or four chamber planes demonstrate the site of the ASD. MR clearly depicts the defect in the portion of the septum separating the superior vena cava from the left atrium (Fig. 8). On spin echo MR images, the thin fossa ovalis can be mistaken for an ASD. To avoid this



Figure 8. A large atrial septal defect is demonstrated both on blackblood turbo-spin echo images (left) and on (dynamic) bright-blood steady-state free-precession images (right). Velocity-encoding MRI was performed in the ascending aorta and the main pulmonary artery, demonstrating significantly increased flow in the MPA (206 mL/s in the MPA vs. 69 mL/s in the ascending aorta). Pulmonary to systemic flow (Qp/Qs) was therefore 3.0.



Figure 9. Gradient-echo cine imaging shows a flow void in the right ventricular outflow tract during systole (arrow), caused by a flow jet through a supracristal ventrical septal defect (VSD). The VSD itself can not be seen in this image as it is not in the imaging plane. LA = left atrium; LV = left ventricle; LVOT = left ventricular outflow tract; Ao = descending aorta.

error the defect should be evident at two adjacent anatomical levels and/or confirmed by a flow jet (void) across the defect on cine MRI.

VEC-MRI measurement in the main pulmonary artery and proximal ascending aorta can be used to calculate pulmonary to systemic flow ratio (Qp/Qs). The imaging planes are placed perpendicular to the direction of blood flow in each artery using sagittal images to site their planes. In the presence of an ASD, Qp/Qs > 1 is found, indicating the hemodynamic significance of the ASD. Good correlations have been found for Qp/Qs, measured by VEC cine MR, and oximetric samples acquired at cardiac catheterization (20, 21).

Ventricular septal defect

MRI in the transaxial horizontal long axis or four chamber planes can precisely demonstrate the site of single or multiple



Figure 10. Flow measurements immediately below the coarctation and at the level of the diaphragm were performed prior to and after intervention in patients with aortic coarctation. The oblique-sagittal black-blood turbo-spin echo MR images show the locations of flow measurements. VEC-MRI was performed prior to and after interventional balloon angioplasty and stent deployment. Before intervention (top graph), flow at the diaphragm level was considerably increased as compared to the more proximal level, demonstrating large (56%) collateral flow. After intervention, normal flow is re-established, the flow at the diaphragm level being approximately 4% less at the more proximal level (bottom graph). The collateral flow has thus been eliminated due to normalized pressure distribution.



VSDs. On cine MR, a flow void apparently passes from the base of the aorta (it actually passes from just below the aortic valve) to the RV outlet region and into the proximal pulmonary artery (Fig. 9). VEC cine MRI can be used to calculate Qp/Qs.

Coarctation of the aorta

MRI is the procedure of non-invasive choice for definitive diagnosis and assessment of the severity of coarctation. MRI

has been shown to be effective for the preoperative assessment of coarctation and for postoperative evaluation of recurrent or persistent hypertension. Spin-echo images in the axial, sagittal, and oblique sagittal planes show morphology of the coarctation and permit measurement of diameter and length of the stenosis.

Gadolinium enhanced 3D MRA can display the entire thoracic aorta on a single image with the use of maximum intensity projection or volume-rendering reconstruction techniques. Gadolinium-enhanced MRA is also effective for demonstrating collateral vessels, which are characteristic for hemodynamically significant coarctation.

Velocity-encoded cine MRI can be applied to demonstrate the presence and estimate the volume of collateral circulation to the descending aorta below the coarctation site. This is accomplished by prescribing (preferably from an oblique sagittal view of the aortic arch) two imaging planes perpendicular to the aorta; one is located about 2 cm beyond the coarctation, and the other at the level of the diaphragm (Fig. 10). In the normal aorta, volume flow is greater (about 5-7% greater) at the proximal site. On the other hand, in presence of a hemodynamically significant coarctation, volume flow is greater at the diaphragm because of retrograde flow through intercostal and mammary arteries and other collaterals into the distal aorta. The presence of greater volume flow at the diaphragmatic level is considered a functional indicator of hemodynamic significance of the coarctation (22, 23). There is a rough linear relationship between percent stenosis and the volume of collateral circulation. After stenting of the coarctation, VEC cine MRI has demonstrated greater volume flow at the proximal site than at the distal one; the collateral flow has thus been reversed and normal flow re-established (Fig. 10). VEC cine can also be used to estimate the gradient across the coarctation. Using a plane perpendicular to the coarctation, peak velocity of flow can be estimated and the peak pressure gradient calculated according to the modified Bernoulli equation, either with a constant (k = 4.0) or with a stenosis-dependent (24) loss coefficient.





Pulmonary arterial anomalies

Pulmonary arterial anomalies are evaluated using spin echo MRI and contrast enhanced 3D MRA for depicting morphology and VEC-MRI for measuring blood flow. VEC-MRI offers the unique capability to assess differential flow in the right and left pulmonary artery in patients with a disparity of blood flow between the right and left pulmonary artery. Such disparity of flow can occur in patients with pulmonary stenosis, hypoplasia, or atresia, or in patients with tetralogy of Fallot. VEC cine MRI is obtained perpendicular to the long axis of the right, left, and main pulmonary arterial segments.

The hemodynamic significance of pulmonary arterial stenoses is assessed by VEC cine MRI measurement of blood flow separately for each pulmonary artery (Fig. 11). Since flow is also measured for the main pulmonary artery, the values can be expressed as percentage of total pulmonary blood flow to each lung. The measurement can be done before and after angioplasty in order to document therapeutic benefit although such measurements may not be possible in the presence of stainless steel stents.

Pulmonary regurgitation

Pulmonic valve insufficiency may be seen as flow jets into the RV outflow tract on SSFP images (Fig. 12). Quantitative assessment of both forward and regurgitant flow is obtained from VEC-MRI in a plane approximately 2 cm distal to the pulmonary valve (25), allowing for calculation of regurgitant fraction (Fig. 12). Alternatively, in the absence of shunts, the difference between right and left ventricular stroke volumes can be assessed from short-axis SSFP images.

Pre- and post-operative evaluation

MRI is an excellent method for both pre- and post-operative evaluation in a variety of congenital heart diseases. Spin-echo images and 3D MRA demonstrate morphology in arbitrary views. Cine images demonstrate valvular function and jets. Moreover, the 3D data set available from cine MR images at multiple phases in the cardiac cycle permit precise quantification of LV and RV volumes, mass, and function. Velocity encoded cine MRI is employed for quantification of volumes and shows left-right disparity, regurgitant fraction, or collateral flow.

Since cine MRI has high interstudy reproducibility, it is the best technique available for serial monitoring of ventricular volumes, mass, and function. This information can be used to decide on optimal timing of surgical or interventional therapy and monitor postoperative success of the therapy. Cine and VEC-MRI have been found to be preferred techniques for monitoring RV function and pulmonary regurgitation in patients with tetralogy of Fallot. RV functional parameters derived from CMR have been used to detect early signs of deterioration of RV function in the presence of severe pulmonary regurgitation (26). Recovery of RV function has been shown with CMR after pulmonary valve replacement (27).

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