Myocardial perfusion imaging with cardiovascular magnetic resonance (CMR) has been established as an accurate method for the detection of coronary artery disease (CAD) in single- and multicenter studies. In most previous studies, myocardial perfusion CMR data have been acquired in 3 left ventricular short-axis slices with an in-plane spatial resolution of 2.5 to 3 mm. In recent years, an array of techniques that can achieve faster scan speed by using spatial, temporal, or spatiotemporal redundancy of the data have been proposed. Applied to myocardial perfusion CMR, the faster data acquisition afforded by these methods has been used in different ways. Some authors have invested the speed-up to improve the in-plane spatial resolution of myocardial perfusion CMR to below 2 mm. The main benefits of such high-resolution acquisition include a reduction in dark-rim artifact and better detection of subendocardial ischemia leading to improved diagnostic performance. Other authors have used faster data acquisition to develop 3-dimensional (3D) myocardial perfusion CMR methods that provide whole-heart coverage. The main motivation for these developments is to overcome the limited cardiac coverage of standard perfusion CMR and to allow more accurate quantification of total myocardial ischemic burden. Furthermore, 3D perfusion CMR allows the acquisition of all slices at the same, optimized time point in the cardiac cycle, for example, the mid-diastolic or end-systolic phase, so that motion artifacts can be reduced and registration between slices improved.

However, both high-resolution and 3D whole-heart perfusion CMR have specific limitations, mostly related to the high-temporal undersampling that is used, and their potential clinical role remains undefined. This article gives a brief overview of the principles of advanced acceleration and compares the current evidence for both high-spatial resolution and 3D whole-heart acquisition in myocardial perfusion CMR.

Advanced Acceleration Methods

Accelerating data acquisition with parallel imaging or echo-planar imaging methods has been standard practice in perfusion CMR for more than a decade, but tradeoffs in signal-to-noise ratio (SNR) and artifacts limit the achievable acceleration to 2- to 3-fold. In a typical stress perfusion study, these methods permit the acquisition of 3 to 4 myocardial slices with an in-plane spatial resolution of 2 to 3 mm. More recently proposed prior-knowledge–based techniques allow much higher acceleration factors for data acquisition. The prior-knowledge principle is not limited to CMR or even MRI, per se, and has also been applied to computed tomography and positron emission tomography. Technical details of current acceleration methods are beyond the scope of this article but can be found in recent technical reviews. Prior-knowledge methods are based on the observation that image data sets exhibit considerable correlation in space and time. Perfusion CMR data sets in particular contain a high degree of temporal redundancy, because data are acquired at a single time point in the cardiac cycle using ECG-gating and during breath-holding, so that most of the image is static and the predominant change between neighboring time frames is related to the relatively slow contrast passage. This image redundancy can be exploited by undersampling data in the time (t) domain in addition to the more conventional undersampling in the spatial (k-space) domain.

Examples of these spatiotemporal (or k-t) undersampling techniques are k-t broad linear acquisition speed-up technique (k-t BLAST), k-t sensitivity encoding (k-t SENSE), and k-t principal component analysis. In k-t BLAST, k-t SENSE, and k-t principal component analysis perfusion CMR sequences, undersampling is applied along k-space and time while a low-spatial resolution image (training data) is obtained in an interleaved fashion during the acquisition. A nonaliased, full image series is then reconstructed using prior-knowledge derived from the training data (Figure I in the online-only Data Supplement). In k-t SENSE and k-t principal component analysis, receiver coil sensitivity information is also used to facilitate image reconstruction. Other related methods have been proposed, such as the highly constrained back-projection reconstruction (HYPRE)
method, in which spatial and temporal redundancy are exploited serially. In HYPR and its variants, k-space data are acquired with undersampled radial projections and overall rotation of the undersampling pattern at different time points. In image reconstruction, a fully sampled composite image is formed by populating missing data from neighboring time frames. This very low-temporal resolution composite is then used to constrain back-projection of the undersampled data acquired for each individual time frame. Other promising acceleration techniques, such as spiral imaging, are in early development, but their clinical utility has yet to be assessed.

With the techniques listed above, it is possible to accelerate image acquisition up to a factor of ≥10 times. This speed-up has been used to improve spatial resolution (high-resolution perfusion CMR) or to acquire 3D myocardial perfusion data within a single acquisition shot facilitating greater spatial coverage.

High-Resolution Perfusion CMR

Although no accepted definition exists, perfusion CMR is considered as high-spatial resolution when the in-plane resolution is better than 2 mm, making the resolution comparable to that of other common CMR methods. The first feasibility study of high-resolution perfusion CMR used 5-fold k-t SENSE to achieve an in-plane spatial resolution of 1.5 mm in a group of 10 volunteers. Image quality was similar to that from a standard-resolution sequence (in-plane spatial resolution of 2.6 mm), but there was a significant reduction in the extent of dark-rim artifact (mean thickness: 1.7 versus 2.4 mm; \( P < 0.01 \)). Images acquired with high-resolution perfusion CMR also displayed an increased SNR compared with the standard-resolution technique when corrected for pixel size, facilitated by constraining the reconstruction with the low-resolution, high-SNR training data. In 3 of the volunteers, higher k-t SENSE acceleration factors of 8 and 10 were successfully used, without compromise in image quality or temporal signal intensity profiles.

The reduction of dark-rim artifact with high-resolution acquisition was confirmed in another volunteer study (\( n=10 \)) by Maredia et al. In this study, mean artifact thickness was 3.4 mm with standard-resolution acquisition compared with only 1.1 mm with a k-t SENSE high-resolution acquisition (\( P < 0.001 \)). Dark-rim artifacts are a common finding in conventional perfusion CMR and are thought to be caused by magnetic susceptibility effects, Gibbs ringing, and cardiac motion during acquisition. Because these artifacts are directly proportional to voxel size, the use of high-resolution perfusion CMR offers a significant advantage.

Following the feasibility studies in volunteers, high-resolution perfusion CMR has been validated in several patient studies (Table 1). In the first of these, Plein et al used an identical sequence to their previous volunteer study (1.5 T, 5-fold k-t SENSE, in-plane resolution 1.4 mm) in 51 patients with known or suspected CAD. High-resolution acquisition was found to have a high-image quality and high-diagnostic accuracy (area under the curve [AUC] = 0.85) against quantitative coronary angiography. Notably, the diagnostic accuracy in single-vessel disease and multivessel disease was similar (AUC: 0.87 versus 0.82, respectively), suggesting that sufficient spatial resolution to resolve a transmural ischemic gradient can overcome one of the major limitations of perfusion imaging in multivessel disease, that is, its dependence on a reference area of normal perfusion (Figure 2). Other benefits of high-resolution acquisition noted in the study were better integration with cine and late-gadolinium enhancement data and minimal dark-rim artifact. A subsequent study demonstrated the clinical feasibility of high-resolution perfusion CMR at 3.0 T and confirmed the expected improvement in image quality and SNR compared with 1.5 T (Table 1).

Manka et al followed with an extension of the k-t SENSE technique at 3.0 T to achieve 8-fold acceleration and an in-plane spatial resolution of 1.1 mm, which was then evaluated in 20 patients with suspected CAD. The combined benefits of higher field strength and greater acceleration led to an incremental improvement in spatial resolution and an even greater AUC of 0.94 (95% confidence interval, 0.74–0.99).

Most recently, in the only study to offer a direct comparison to date, high-resolution perfusion CMR (8-fold k-t BLAST, in-plane spatial resolution 1.6 mm) was found to have a significantly greater overall diagnostic accuracy compared with standard-resolution technique in 100 patients with suspected CAD (AUC, 0.93 versus 0.83; \( P < 0.001 \)). The improved diagnostic performance was attributed to better detection of subendocardial ischemia (Figure 2). Because the endocardial layer is the most vulnerable to ischemia, the ability to detect subendocardial perfusion deficits confidently can be expected to improve the detection of CAD in a perfusion study. Furthermore, a recent study by Hautvast et al. showed that high-resolution perfusion CMR data can be used to quantify transmural perfusion gradients accurately, and these new measurements may serve as diagnostic markers for the detection and characterization of epicardial coronary disease as well as microvascular disease.

The HYPR method was used in 1 recent single-center study, in a variant known as sliding-window conjugate-gradient HYPR (6 contiguous slices, 1.6 mm in-plane
spatial resolution). The study demonstrated clinical feasibility and a high-diagnostic accuracy in 50 patients with suspected CAD.25

Whereas the previous studies have all used quantitative coronary angiography as their reference standard, Lockie et al11 validated high-resolution perfusion CMR against invasive pressure-wire–derived fractional flow reserve (FFR). In the 42 patients studied, high-resolution perfusion CMR (5-fold $k$-t BLAST, in-plane resolution 1.2 mm) had a high diagnostic accuracy (AUC=0.92) for the detection of hemodynamically significant lesions (FFR<0.75). To have a noninvasive method that so closely correlates with FFR is highly attractive, and more widespread use could have a significant impact on clinical pathways, especially in patients with anatomically complex and multivessel disease.32 However, whether high-resolution perfusion CMR conveys a diagnostic increment against FFR is not certain, and notably in a previous study using FFR<0.75 as the reference standard, Watkins et al33 found a high level of diagnostic performance with a standard-resolution acquisition (sensitivity 91%, specificity 94%), comparable with the high-resolution method used by Lockie et al.11

Three-Dimensional Whole-Heart Perfusion CMR

Conventionally, cardiac coverage with perfusion CMR is limited to 3 to 4 noncontiguous slices through the left ventricle. However, nuclear perfusion studies have demonstrated that the extent of hypoperfusion and overall ischemic burden is a strong marker of clinical outcome.34,35 It is unknown how accurately ischemic burden can be measured from noncontiguous image sections. Therefore, despite the demonstrated high diagnostic accuracy of conventional 2D perfusion CMR, the lack of complete myocardial coverage remains a potential limitation of the method.

The same speed-up methods that permit high-resolution myocardial perfusion CMR can also be used to acquire a single 3D stack, covering all or a large proportion of the heart while preserving adequate temporal and spatial resolution (Figure 3).2 Advanced acceleration methods allow the acquisition of 3D data covering the whole heart in up to

Table 1. Diagnostic Performance of Advanced Accelerated Perfusion CMR Techniques

<table>
<thead>
<tr>
<th>Study</th>
<th>N</th>
<th>Acceleration Method</th>
<th>In-Plane Resolution</th>
<th>Spatial Coverage</th>
<th>Reference Standard</th>
<th>Diagnostic Accuracy</th>
</tr>
</thead>
<tbody>
<tr>
<td>High-Resolution</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Plein 20088</td>
<td>51</td>
<td>1.5 T 5× k-t SENSE</td>
<td>1.4 mm</td>
<td>4 slices (NC)*</td>
<td>QCA&gt;50%</td>
<td>0.85†</td>
</tr>
<tr>
<td>Plein 20089</td>
<td>33</td>
<td>1.5 T 5× k-t SENSE</td>
<td>1.5 mm</td>
<td>4 slices (NC)*</td>
<td>QCA&gt;50%</td>
<td>0.80†</td>
</tr>
<tr>
<td>Plein 20089</td>
<td>33</td>
<td>3.0 T 5× k-t SENSE</td>
<td>1.3 mm</td>
<td>4 slices (NC)*</td>
<td>QCA&gt;50%</td>
<td>0.89†</td>
</tr>
<tr>
<td>Manka 201012</td>
<td>20</td>
<td>3.0 T 8× k-t SENSE</td>
<td>1.1 mm</td>
<td>3 slices (NC)</td>
<td>QCA&gt;50%</td>
<td>0.94†</td>
</tr>
<tr>
<td>Motwani 201212</td>
<td>100</td>
<td>1.5 T 8× k-t BLAST</td>
<td>1.6 mm</td>
<td>3 slices (NC)</td>
<td>QCA≥50%</td>
<td>0.93†</td>
</tr>
<tr>
<td>Lockie 201111</td>
<td>42</td>
<td>3.0 T 5× k-t BLAST</td>
<td>1.2 mm</td>
<td>3 slices (NC)</td>
<td>FFR&lt;0.75</td>
<td>0.92†</td>
</tr>
<tr>
<td>3-Dimensional</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Manka 201115</td>
<td>146</td>
<td>3.0 T 6× k-t SENSE</td>
<td>2.3 mm</td>
<td>16 slices (WH)</td>
<td>QCA≥50%</td>
<td>0.83‡</td>
</tr>
<tr>
<td>Manka 201216</td>
<td>120</td>
<td>1.5 T 10× k-t PCA</td>
<td>2.0 mm</td>
<td>16 slices (WH)</td>
<td>FFR&lt;0.75</td>
<td>0.87‡</td>
</tr>
<tr>
<td>Jogiya 201217</td>
<td>53</td>
<td>3.0 T 10× k-t PCA</td>
<td>2.3 mm</td>
<td>12 slices (WH)</td>
<td>FFR&lt;0.75</td>
<td>0.91‡</td>
</tr>
<tr>
<td>HYPR Ma 201225</td>
<td>50</td>
<td>3.0 T SW-CG-HYPR</td>
<td>1.6 mm</td>
<td>6 slices (WH)</td>
<td>QCA≥50%</td>
<td>0.90‡</td>
</tr>
</tbody>
</table>

BLAST indicates broad linear speed-up technique; CMR, cardiovascular magnetic resonance; FFR, fractional flow reserve; HYPR, highly constrained back-projection reconstruction; NC, noncontiguous; PCA, principal component analysis; QCA, quantitative coronary angiography; SENSE, sensitivity encoding; SW-CG-HYPR, sliding-window conjugate-gradient HYPR; T, Tesla; and WH, whole-heart.

*In these studies, temporal resolution was 2 R-R intervals, whereas in all others, it was 1 R-R interval.
†In these studies, diagnostic accuracy was calculated by receiver operating characteristic analysis and expressed as area under the curve.
‡In these studies, diagnostic accuracy was expressed as the proportion of correctly classified subjects (true positives+true negatives) among all subjects.

Figure 2. High-resolution perfusion cardiovascular magnetic resonance (CMR) in multivessel disease. A 48-year-old woman with suspected angina underwent stress perfusion CMR at 1.5 T using a high-resolution (1.4 mm in-plane) technique facilitated by 5-fold k-t broad linear speed-up technique. Stress-induced perfusion defects were seen in all 3 territories (arrows; A and B) and are clearly demarcated as subendocardial. Subsequent x-ray angiography revealed significant 3-vessel disease (arrows; C and D). This case highlights the ability of high-resolution acquisition to overcome the potential effects of balanced ischemia in multivessel disease by detecting transmural perfusion gradients and subendocardial ischemia.

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16 contiguous slices within an acquisition duration of <200 ms. The most significant advantage of 3D acquisition is that contiguous spatial coverage allows the true extent of perfusion defects to be visualized. Furthermore, although data in the literature regarding the incidence of pure apical ischemia are lacking and they are presumed to be rare, the inability of 2D perfusion CMR to image the apical cap is sometimes stated as a limitation, and this is overcome by 3D acquisition.

Another limitation of conventional 2D-myocardial perfusion CMR is that each slice is acquired in a different phase of the cardiac cycle (Figure 3). In practice, this means that the acquisition cannot be optimized for all slices and may coincide with rapid cardiac motion in some slices, leading to motion artifacts in the images. In 3D acquisition, all data are acquired in the same cardiac phase, and the time point for acquisition in the cardiac cycle can be optimized to minimize artifact. Furthermore, the cardiac phase can be matched with other acquired data, such as late-gadolinium enhancement imaging for improved coregistration. As a related benefit, 3D acquisition also permits the optimal choice of the saturation recovery time and the ability to scan at high-heart rates, because only 1 preparation pulse and 1 readout per heart beat are applied. The optimal phase within the cardiac cycle for 3D acquisition is the subject of ongoing research, with end-systole and mid-diastole being the favored options. In mid-diastole, the heart is usually at its most stationary, but the relatively thin myocardium can compound the effect of dark-rim artifact and limit the assessment of the transmurality perfusion defects. End-systole, on the other hand, has a shorter quiescent period, but the thicker myocardium reduces the effect of dark-rim artifact and facilitates the grading of defect transmurality. Systolic acquisition is also less sensitive to R-R variability and arrhythmia, which is relevant to the increasing burden of patients with atrial fibrillation. Notably, recent studies have found significant differences between systolic and diastolic myocardial blood flow estimates with perfusion CMR, and the choice of either cardiac phase for 3D myocardial perfusion may, therefore, be of physiological interest or relevant to certain disease processes, such as hypertrophic cardiomyopathy.

Therefore, 3D myocardial perfusion CMR is a highly promising development, and recent studies have shown it to be clinically feasible, highly accurate, and to have a potential role in the assessment and follow-up of ischemic burden. In the first clinical study of 3D perfusion CMR, Manka et al evaluated 146 patients with suspected CAD using $k-t$ acceleration to achieve 16-slice coverage and an

Figure 3. Pulse sequence diagrams for 2-dimensional (2D) and 3D perfusion cardiovascular magnetic resonance (CMR). In 2D perfusion CMR (A), 3 to 4 noncontiguous slices are acquired in different phases of the cardiac cycle to maximize spatial coverage. Each slice is acquired after a saturation prepulse (black bar). A shared prepulse can be used to save time but results in different contrast characteristics for each slice. In 3D perfusion CMR (B), a single saturation prepulse is followed by a saturation recovery time ($T_{SR}$) and undersampled 3D perfusion data readout. The use of advanced spatiotemporal undersampling allows sufficient data acquisition to reconstruct 12 to 16 contiguous slices, that is, whole-heart coverage. The training data are acquired with the undersampled data in an interleaved fashion. Because all perfusion data in the 3D technique are acquired at the same point in the cardiac cycle, the reconstructed slices all appear in the same cardiac phase.
in-plane spatial resolution of 2.3 mm. Image quality was consistently high, and the overall sensitivity, specificity, and diagnostic accuracy to detect significant CAD (quantitative coronary angiography ≥50%) were 92%, 74%, and 83%, respectively, comparable with conventional 2D perfusion CMR.1 Furthermore, myocardial ischemic burden (expressed as a percentage of left ventricle myocardial hypoenhancement to total left ventricle myocardium) was calculated from 3D perfusion CMR data (Figure 4). In 48 patients who had a repeat scan after PCI, there was a relative reduction in myocardial ischemic burden of 79±25%, highlighting the potential role of 3D perfusion CMR to monitor the response to anti-ischemic therapies serially.15 In addition, the quantification of ischemic burden using myocardial hypoenhancement volumetry (with an arbitrary signal intensity threshold of 2 standard deviations below remote myocardium) was found to be highly reproducible on intrareader and inter-reader assessment.

Although the previous study used quantitative coronary angiography ≥50% as the reference standard for CAD, 2 recent studies (1 at 1.5 T and 1 at 3.0 T) have validated 3D perfusion CMR against FFR and shown high-diagnostic accuracy of the method.16,17 In a study by Manka et al,16 3D perfusion CMR at 1.5 T was found to have a sensitivity, specificity, and diagnostic accuracy of 90%, 82%, and 87%, respectively. At 3.0 T, a study by Jogiya et al17 found similar figures of 91%, 90%, and 91%, respectively. Both of these studies further verified the feasibility and reproducibility of ischemic burden quantification using volumetry of myocardial hypoenhancement. In addition, Manka et al also found that the quantified myocardial ischemic burden had a high-diagnostic accuracy (AUC=0.90) for identifying FFR-defined CAD, with an optimal cutoff value of >4.4%.16

Although FFR is widely considered the invasive reference standard for determining the hemodynamic significance of a coronary stenosis, it does not provide any information about the magnitude of consequent ischemia. This deficiency highlights a potential role for 3D perfusion CMR, and in both of the validation studies against FFR, it demonstrated a higher ischemic burden with proximal lesions compared with distal lesions.16,17 Furthermore, Jogiya et al17 found a strong correlation (r=0.82; 95% confidence interval, 0.70–0.89; P<0.0001) between the myocardial ischemic burden on 3D perfusion CMR and the Duke Jeopardy score, which is a validated invasive assessment of ischemic burden based on lesion severity and location.

Figure 4. Myocardial ischemic burden quantification with 3-dimensional (3D) perfusion cardiovascular magnetic resonance (CMR). A, Shows consecutive slices of a 3D perfusion CMR scan during adenosine stress in a patient with significant stenoses in the proximal right coronary artery and distal left anterior descending coronary artery. B, Shows identical images illustrating volumetry of myocardial hypoenhancement using a signal intensity threshold of 2 SDs below remote myocardium (red areas). The volume of myocardial hypoenhancement was 20.4% of total myocardium.
Limitations of Highly Accelerated Perfusion CMR

$k-t$ techniques add complexity to the acquisition of perfusion CMR, and they are sensitive to respiratory motion and cardiac arrhythmias. To reduce respiration-related artifacts when using these techniques, fewer dynamic images tend to be acquired compared with conventional methods, and more emphasis is placed on respiratory coaching. To reduce respiration-related artifacts when using these techniques, fewer dynamic images tend to be acquired compared with conventional methods, and more emphasis is placed on respiratory coaching. The acquisition of fewer dynamic images does, however, confer a risk of inadequate temporal sampling of myocardial contrast passage, particularly if there is significant arrhythmia, and therefore patients with arrhythmia may not be ideal for these techniques. In practice, the breath-hold capacity of patients can be maximized by clear breathing instructions, trial runs, and acquisition in inspiration. Alternative strategies are shallow respiration throughout the acquisition or a more focused breath-hold during myocardial contrast passage.

Previous studies have shown that in compliant patients, respiratory artifacts are rare and tend to occur at the end of a breath-hold; if the timing of the acquisition is correctly synchronized to the breath-hold command, such artifacts rarely interfere with image interpretation. Furthermore, recent improvements, such as $k-t$ principal component analysis, are less sensitive to respiratory motion and improve temporal fidelity (compared with $k-t$ SENSE and $k-t$ BLAST reconstructions that can suffer from temporal blurring), because image reconstruction is constrained using temporal basis functions derived from the low-resolution training data acquired in every heartbeat. Advanced motion correction techniques that allow free-breathing throughout acquisition are also in development.

A further limitation is that the described reconstruction techniques all assume the heart remains entirely static, but there are of course small movements throughout the cardiac cycle that can cause motion-related artifacts, including dark-rim artifact. 3D acquisition is more vulnerable to motion-related artifact than 2D acquisition, as well as respiratory artifact, because of its larger temporal footprint (length of acquisition is $\approx 200$ ms for 3D compared with $<100$ ms for most 2D acquisitions).

Finally, and this applies to all undersampling methods, reducing the number of sampled data points leads to an SNR reduction that is proportional to $1/\sqrt{R}$, where $R$ is the acceleration factor. In $k-t$ methods, the actual SNR reduction may, however, be much less if the dynamics of the object are highly correlated in successive time frames.

High-Spatial Resolution or Full Cardiac Coverage?

Whether the benefits of high-spatial resolution, such as detection of subendocardial ischemia and reduction of dark-rim artifact, outweigh the potential benefits of whole-heart coverage, including more reliable ischemia quantification, is a complex and as yet unanswered question. Initial studies, albeit in selected and small patient populations, suggest that the diagnostic accuracy of the 2 approaches is similar, but there has not yet been a direct head-to-head comparison, and further large-scale studies are required. In clinical practice,
the 2 different approaches may have specific benefits for individual patients. In a patient with a de novo suspicion of CAD, the available limited data on diagnostic accuracy suggest that either technique, or indeed standard-resolution perfusion CMR, would be appropriate choices. However, in a patient who has already undergone angiography and is found to have diffuse multivessel disease with possible PCI targets, high-resolution acquisition might be favorable on account of its ability to detect subendocardial ischemia in the presence of balanced ischemia (Figure 2). In patients with a history of myocardial infarction and suspected peri-infarct ischemia, a 3D perfusion scan may better allow matching of late-gadolinium enhancement and perfusion images over the entire heart with more reliable quantification of peri-infarct ischemia, rather than an assumption based on 3 sparse slices. Figures 5–8 show examples of patients undergoing both acquisitions, and the relative merits (Table 2) of both techniques are discussed in each case.

An alternative strategy to either high-resolution or whole-heart coverage would be to divide the speed-up afforded by acceleration techniques between spatial resolution and cardiac coverage. However, at present, there is insufficient data to define the optimal compromise between these parameters adequately. The subendocardial layer is the most vulnerable to ischemia, and therefore sufficient spatial resolution to resolve a transmural perfusion gradient is preferable, and this remains a significant advantage of CMR over single-photon emission computed tomography (SPECT). However, it is unclear how far this notion needs to be taken or whether the spatial resolution used in recent large-scale studies is sufficient, particularly given the high levels of diagnostic performance. Only a large-scale clinical study comparing acquisitions of varying spatial resolution and differing degrees of cardiac coverage can answer this question. Furthermore, whether an optimal compromise is best acquired throughout the cardiac cycle with a multislice 2D approach, en bloc with a 3D approach or with newer techniques, such as HYPR, would also have to be determined. Finally, beyond studies of diagnostic accuracy, we also need to evaluate whether the various approaches differ in their quantification of ischemic burden, particularly given the increasing use of CMR for this purpose in clinical studies and for accurate prognostic information.

In this review, we have deliberately focused on the visual analysis of perfusion CMR, because it remains the most common method of interpretation in clinical practice. However, quantitative methods for the estimation of myocardial blood flow from conventional myocardial perfusion CMR data have been clinically validated. We have shown in previous studies that quantitative analysis of high-resolution perfusion data acquired with high spatiotemporal undersampling methods is also feasible and offers additional intriguing opportunities, such as quantitation of transmural perfusion gradients. However, the algorithms applied for the reconstruction of high-resolution and 3D perfusion CMR data acquired with spatiotemporal undersampling methods give rise to a degree of low-pass temporal filtering, posing additional challenges for quantitative assessment. Quantitative analysis of 3D perfusion

Figure 6. Case example 2. A 45-year-old man with previous PCI to the LAD represented with significant angina. The top panel dimensionally shows 3-dimensional (3D) perfusion cardiovascular magnetic resonance (CMR; 12 slices) at stress; the middle panel shows late-gadolinium enhancement (LGE) imaging; and the bottom panel shows high-resolution (1.1 mm in-plane) stress perfusion CMR, all performed at 3.0 T. 3D perfusion CMR shows stress-induced hypoperfusion throughout the anterior wall from base to apex, that is, well beyond the area of scar seen in the mid-anterior wall on LGE imaging. This example shows the benefit of whole-heart coverage with the 3D acquisition, because the 3-slice high-resolution techniques did not demonstrate any significant ischemia beyond the established scar in the mid-ventricle. X-ray angiography confirmed a subtotal occlusion of a large diagonal branch, accounting for the anterior ischemia (black arrow). LAD indicates left anterior descending artery; PCI, percutaneous coronary intervention; and VLA, vertical long axis.
data has not yet been reported, and therefore a comparison of myocardial blood flow estimates from standard-resolution, high-resolution, and 3D whole-heart perfusion data must be reserved for a future discussion.

**Conclusion**

The application of advanced acceleration techniques in perfusion CMR has led to 2 highly accurate alternative strategies with distinct advantages. Based on current evidence, it is not clear whether high-resolution or 3D whole-heart coverage offers greater clinical benefit, and large comparative studies are needed to address this question. However, where expertise in both exist, the choice between high-resolution and 3D perfusion CMR already gives us the opportunity to tailor the type of perfusion sequence used to an individual patient to best answer specific clinical questions. Finally, future developments may lead to an acquisition with the optimal balance of spatial resolution and cardiac coverage, but further research is first needed to determine the minimum number of slices required for accurate estimation of ischemic burden and the
We thank Gavin Bainbridge, Margaret Saysell, and Caroline level of spatial resolution required to optimally detect clinically significant ischemia.

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**Key Words:** cardiac magnetic resonance imaging ■ ischemic heart disease ■ perfusion imaging
Advanced Cardiovascular Magnetic Resonance Myocardial Perfusion Imaging: High-Spatial Resolution Versus 3-Dimensional Whole-Heart Coverage
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The volume of myocardial hypoenhancement in Figure 4 appeared incorrectly in the final sentence of the figure legend on page 343. The sentence has been corrected and now appears as follows:

“The volume of myocardial hypoenhancement was 20.4% of total myocardium.”

These errors have been corrected in the online version of the article, which is available at: http://circimaging.ahajournals.org/content/6/2/339.full.
Supplemental Figure I

(a) In $k$-t BLAST and $k$-t SENSE, data acquisition is accelerated by under-sampling along the spatial encoding ($k$) and time ($t$) axes which leads to an aliased image. (b) Low resolution ‘training data’ is also obtained to determine data correlations i.e. ‘prior-knowledge’. (c) Finally, a non-aliased full image series is reconstructed using a statistical model derived from the training data. BLAST= broad linear speed-up technique; SENSE=sensitivity encoding.