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Dear MAGNETOM Flash reader,

One could easily argue that cardiovascular magnetic resonance (CMR) is in the midst of another technical revolution. Those of us who have worked in the field for the last two decades have seen similar periods in the past, when major advances in hardware technology like array coils and fast gradients, in software technology like parallel acquisition techniques, and pulse sequences like balanced steady-state free precession have spawned dramatic improvements in the efficiency and effectiveness of CMR. Three of the most exciting recent advances: myocardial parameter mapping, MR-PET, and compressed sensing are highlighted in the six articles of this issue of MAGNETOM Flash. Relaxation parameter mapping has initiated an exciting new direction of research into the clinical implications of diffuse changes in myocardial tissue (e.g., fibrosis or edema) that can accompany a variety of diseases. The novel combination of CMR and PET is enabling the powerful diagnostic combination of the exquisite assessment of myocardial tissue structure and function provided by CMR, together with the evaluation of metabolism by PET. New approaches to sampling and reconstruction using compressed sensing are dramatically reducing the data acquisition requirements, and thereby significantly enhancing the efficiency of CMR. Together, these advances are indicative of the ever-changing nature of CMR, a technology that continues to improve thanks to the passion, creativity, and tireless effort of the researchers around the world who have made this their life’s work.

The article by Moon et al., from University College London Hospitals, London, UK, discusses recent trends in the development and investigation of techniques for quantitative mapping of myocardial T1 and T2 relaxation parameters. Quantitative mapping addresses many of the technical limitations of conventional T1-weighted and T2-weighted sequences, most importantly offering the capability to assess diffuse changes in myocardial tissue that can accompany many disease states. The article by Fernandes et al., from University of Campinas, Brazil, nicely summarizes the techniques that are employed for myocardial T1 mapping, and reviews recent investigations of this technology in patients with a variety of diseases including amyloid, aortic stenosis, and various cardiomyopathies. As noted in both articles, the early evidence indicates that myocardial relaxation parameter mapping has fantastic potential as a diagnostic tool that may be sensitive to early pathological changes in myocardial tissue potentially missed by other imaging methods. Challenges remain in standardization of these methods to ensure consistent quantitative results across patients and imaging platforms.

The article by Schwitter et al., from the Cardiac Magnetic Resonance Center of the University Hospital of Lausanne in Switzerland nicely demonstrates an important advantage of highly accelerated cine imaging using compressed sensing data acquisition and reconstruction strategies. The ability to acquire sufficient cine slices to cover the entire heart in multiple orientations in a single breath-hold (2 beats per slice) not only reduces exam times, but also facilitates more accurate LV volume calculations using a three dimensional modeling approach rather than the traditional Simpson’s Method. Reducing the potential for mis-registration of slices avoids one of the primary limitations of the 3D approach to LV volume calculations. Thus, the efficiency gains achieved via compressed sensing data acquisition and reconstruction strategies can positively impact the clinical value of CMR from several different perspectives.

The article by Carr et al., from the group at Northwestern University in Chicago highlights the tremendous potential of iterative reconstruction techniques to dramatically accelerate cardiac imaging. The results shown indicate that efficiency gains of at least a factor of two are possible over conventional parallel acquisition techniques. The time-consuming nature of most CMR techniques, and the requirements of fasted patients, breath-holds, and regular cardiac rhythm are factors that have constrained the widespread acceptance of CMR into the clinical routine. While there is still work remaining to optimize data sampling and reduce image reconstruction times, the gains in scanning efficiency demonstrated in this study could have far-reaching implications in moving CMR further into the mainstream as a cost-effective diagnostic imaging modality.

The potential advantages of simultaneous CMR and PET acquisitions are explored in two articles of this issue of MAGNETOM Flash. Drs. Cho and Kong from Yeungnam University Hospital, Daegu, South Korea, demonstrate in a patient with hypertrophic cardiomyopathy the ability to characterize myocardial fibrosis using both Late Gadolinium Enhancement and 18F-FDG PET.

The article by Dr. James A. White from The Lawson Health Research Institute, London, Ontario, Canada, nicely describes the potential for advanced myocardial tissue characterization using the synergistic capabilities of CMR and PET. Dr. White points out how the complimentary and unique information provided by CMR and PET may better characterize pathological changes in myocardial tissue in diseases such as sarcoidosis. The evaluation of cellular metabolic activity using PET may fill the role that MR spectroscopy has promised but as yet been unable to deliver in the clinical setting. The field of metabolic imaging is rapidly evolving.

Orlando P. Simonetti, Ph.D.
The entire editorial staff at The Ohio State University and at Siemens Healthcare extends their appreciation to all the radiologists, technologists, physicists, experts and scholars who donate their time and energy – without payment – in order to share their expertise with the readers of MAGNETOM Flash.
New Generation Cardiac Parametric Mapping: the Clinical Role of T1 and T2 Mapping

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Introduction
Cardiovascular magnetic resonance (CMR) is an essential tool in cardiology and excellent for cardiac function and perfusion. However, a key, unique advantage is its ability to directly scrutinize the fundamental material properties of myocardium—‘myocardial tissue characterization’. Between 2001 and 2011, the key methods for tissue characterization have been sequences ‘weighted’ to a magnetic property—T1-weighted imaging for scar (LGE) and T2-weighted for edema (area at risk, myocarditis). These, particularly LGE imaging, have changed our understanding and clinical practice in cardiology.

However, there are limitations to these approaches: Both are difficult to quantify in non-ischemic disease. These techniques exhibit a dichotomy of health and disease. As a result, global myocardial pathologies such as diffuse infiltration (fibrosis, amyloid, iron, fat, pan-inflammation) are missed.

Recently, rapid technical innovations have generated new ‘mapping’ techniques. Rather than being ‘weighted’, these create a pixel map where each pixel value is the T1 or T2 (or T2*) displayed in color. These new sequences are single breath-hold, increasingly robust and now widely available. With T1 mapping, clever contrast agent use also permits the measurement of the extracellular volume (ECV). The ECV is important. A multiparametric approach (e.g. T2 mapping or T2-weighted imaging in addition) may therefore be useful. Amyloid can have far higher ECVs than any other disease [18] whereas ageing has small changes—near the detection limits, but of high potential clinical importance [19, 20]. For low ECV expansion diseases, biases from blood pool partial volume errors need to be meticulous addressed. Nevertheless, even modest ECV changes appear prognostic. In 793 consecutive patients (all-comers but excluding amyloid and HCM, measuring outside LGE areas) followed over 1 year, global ECV predicted short-term mortality (Fig. 2).

T1 mapping
Initial T1 measurement methods were multi-breath-hold. These were time consuming and clunky, but were able to measure well diffuse myocardial fibrosis, a fundamental myocardial property with high potential clinical significance [1]. Healthy volunteers and those with disease had different extents of diffuse fibrosis [2], and these were shown to be clinically significant in a number of diseases.

Two key ways of using T1 mapping: Without (or before) contrast—Native T1 mapping; and with contrast, typically by subtracting the pre and post maps with hematocrit correction to generate the ECV [6].

Native T1
Native T1 mapping (pre-contrast T1) can demonstrate intrinsic myocardial contrast (Fig. 1). T1, measured in milliseconds, is higher where the extracellular compartment is increased. Fibrosis (focal, as in infarction, or diffuse) [7-8], edema [9-10] and amyloid [11], are examples. T1 is lower in lipid (Anderson Fabry disease, AFD) [12], and iron [13] accumulation. These changes are large in some rare disease. Global myocardial changes are robustly detectable without contrast, even in early disease. In iron, AFD and amyloid, changes appear before any other abnormality—they may be no left ventricular hypertrophy, a normal electrocardiogram, and normal conventional CMR, for example—generally new information. In established disease, low T1 values in AFD appear to absolutely distinguish it from other causes of left ventricular hypertrophy, a near detection limit, whilst in established amyloid T1 elevation tracks known markers of cardiac severity [11].

A note of caution, however. Native T1, although stable between healthy volunteers to 1 part in 30, is dependent on platform (magnet manufacturer, sequence and sequence variant, field strength) [14]. Normal reference ranges for your setup are needed.

The signal acquired is also a composite signal—generated by both interstitium and myocytes. The use of an extracellular contrast agent adds another dimension to T1 mapping and the ability to characterize the extracellular compartment specifically.

Extracellular volume (ECV)
Initially, post-contrast T1 was measured, but this is confounded by renal clearance, gadolinium dose, body composition, acquisition time post bolus, and hematocrit. Better is measuring the ECV. The ratio of change of T1 between blood and myocardium after contrast, at sufficient equilibrium (e.g. after 15 minutes post-bolus—no infusion generally needed) [15, 16], represents the contrast agent partition coefficient [17], and if corrected for the hematocrit, the myocardial extracellular space—ECV [1]. The ECV is specific for extracellular expansion, and well validated. Clinically this occurs in fibrosis, amyloid and edema. To distinguish, the degree of ECV change and the clinical context is important. A multiparametric approach (e.g. T2 mapping or T2-weighted imaging in addition) may therefore be useful. Amyloid can have far higher ECVs than any other disease [18] whereas ageing has small changes—near the detection limits, but of high potential clinical importance [19, 20]. For low ECV expansion diseases, biases from blood pool partial volume errors need to be meticulously addressed. Nevertheless, even modest ECV changes appear prognostic. In 793 consecutive patients (all-comers but excluding amyloid and HCM, measuring outside LGE areas) followed over 1 year, global ECV predicted short-term mortality (Fig. 2).
Progress is rapid, challenges remain. Delivery across sites and standardization is now beginning with new draft guidelines for T1 mapping in preparation. Watch this space.

References
Myocardial T1 Mapping: Techniques and Clinical Applications

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Introduction
Cardiovascular magnetic resonance (CMR) has been an increasingly used imaging modality which has experienced significant advancements in the last years [1]. One of the most used techniques that have made CMR so important is late gadolinium enhancement (LGE) and the demonstration of localized areas of infarct and scar tissue [2–4]. However, despite being very sensitive to small areas of regional fibrosis, LGE techniques are mostly dependent on the comparison to supposedly normal reference areas of myocardium, thus not being able to depict more diffuse disease.

Myocardial interstitial fibrosis, with a diffuse increase in collagen content in myocardial volume, develops as a result of many different stimuli includ-

Table 1: Comparison of the MOLLI sequences available for T1 mapping

<table>
<thead>
<tr>
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<tbody>
<tr>
<td>Preparation</td>
<td>Non-selective inversion recovery</td>
<td>Non-selective inversion recovery</td>
<td>Non-selective inversion recovery</td>
</tr>
<tr>
<td>Bandwidth</td>
<td>1090 Hz/px</td>
<td>1090 Hz/px</td>
<td>1090 Hz/px</td>
</tr>
<tr>
<td>Flip angle</td>
<td>50°</td>
<td>35°</td>
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</tr>
<tr>
<td>Base matrix</td>
<td>240</td>
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<tr>
<td>Phase resolution</td>
<td>151</td>
<td>128</td>
<td>144</td>
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<tr>
<td>FOV × % phase</td>
<td>380 × 342</td>
<td>256 × 100</td>
<td>340 × 75</td>
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<tr>
<td>T1</td>
<td>100 ms</td>
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<td>100 ms</td>
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<tr>
<td>Slice thickness</td>
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<td>8 mm</td>
</tr>
<tr>
<td>Acquisition window</td>
<td>191.1 ms</td>
<td>202 ms</td>
<td>206 ms</td>
</tr>
<tr>
<td>Trigger delay</td>
<td>300 ms</td>
<td>300 ms</td>
<td>500 ms</td>
</tr>
<tr>
<td>Inversions</td>
<td>3</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Acquisition heartbeats</td>
<td>3,3,5</td>
<td>3,3,5</td>
<td>5,5,1</td>
</tr>
<tr>
<td>Recovery heartbeats</td>
<td>3,3,1</td>
<td>3,3,1</td>
<td>1,1,1</td>
</tr>
<tr>
<td>TI increment</td>
<td>100–150 ms</td>
<td>80 ms</td>
<td>80 ms</td>
</tr>
<tr>
<td>Scan time</td>
<td>17 heartbeats</td>
<td>17 heartbeats</td>
<td>9 heartbeats</td>
</tr>
<tr>
<td>Spatial resolution</td>
<td>2.26 × 1.58 × 8 mm</td>
<td>2.1 × 1.8 × 8 mm</td>
<td>1.8 × 1.8 × 8 mm</td>
</tr>
</tbody>
</table>

MOLLI images (1A) with respective signal-time curves (1B) and reconstructed T1 map (1C) at 3T. The mean T1 time for this patient was 1152 ms (pre-contrast).

1A
1B
1C
while equilibrium contrast definitely an area where CMR plays myocardial tissue characterization is be more suitable. Therefore, if one wants to image injury, LGE has a limited sensitivity deposits appear in areas of myocyte fibrosis, where regional collagen content \[5\]. Different from replacement renin-angiotensin-aldosterone sys- tems. Another implementation of TrueFISP \[12, 13\] or multishot saturation-recovery images \[14\] but their reproducibility and accuracy have not been extensively validated. The most widely used T1 mapping sequence is based on the Modified Look-Locker Inversion-recovery (MOLLI) technique. Described originally by Messroghli et al. \[15\] it consists of a single shot TrueFISP image with acqui- sitions over different inversion time readouts allowing for magnetization recovery of a few seconds after 3 to 5 readouts. The parameters for the original MOLLI sequence are described in Table 1. The advantages of this sequence over previous methods are its acquisition in only one relatively short breath-hold, the higher spatial resolution (1.6 × 2.3 × 8 mm) and increased dynamic signal. Reproduc- ibility studies using this sequence have shown that the method is very accu- rate with a coefficient of variation of 5.4% \[16\] although an underestima- tion of 8% should be expected based on phantom data. An example of MOLLI images and its respective signal-time curves and map are shown in Figure 1. One disadvantage of this implementa- tion of MOLLI is its dependence on heart rate, mostly true for T1 values less than 200 msec or greater than 750 msec. However, because the devi- ation is systematic, raw values can be corrected using the formula T1 \(\text{corrected} = T1_{\text{raw}} - (2.7 \times \text{[heart rate - 70]})\), bringing the coefficient of variation down to 4.6% after applying the correction. An optimized MOLLI sequence was subsequently described where heart rate correction might not be even nec- essary \[17\]. In the optimized sequence, the authors tested variations in readout flip angle, minimum inversion time, inversion time increments and number of pauses between each readout sequence. The conclusion from these experiments showed that a flip angle of 35°, a minimum inversion time of 100 msec, increments of 80 msec and three heart cycle pauses allowed for the most accurate measurement of myocardial T1 (Table 1). Because T1 assessment may be sensitive to motion artifacts and not all patients might be able to hold their breaths through- out all the necessary cardiac cycles used in MOLLI’s sequence implementa- tion, more recently a shortened version sequence (ShMOLLI) using only 9 heart beats was presented to account for those limitations \[18\]. Using incomplete recovery of the lon- gitudinal magnetization that is cor- rected directly in the scanner by conditional interpretation, ShMOLLI was directly compared to MOLLI in patients over a wide range of T1 times and heart rates both at 1.5 and 3T. The results showed that despite an increase in noise and slight increase in the coefficient of variation (espe- cially at 1.5T), T1 times were not sig- nificantly different using ShMOLLI with the advantage of much shorter acquisition times (9 ± 1.1 sec versus 17.6 ± 2.9 sec). An example of MOLLI and ShMOLLI images from the same patient is presented in Figure 2. Up to now, after acquiring images for T1 mapping, one had to analyze them using in-house developed soft- ware, dedicated commercial pro- grams or open-source solutions \[19\], not always a simple and routine task, leading to difficulty in post-process- ing the data and generating T1 values. Recent advances have provided new- inline processing techniques that will generate the T1 maps automatically after image acquisition with MOLLI, without the need for further post-pro- cessing, accelerating the whole pro- cess. An example of such automated T1 map is presented in Figure 3. At the same time, online application of motion correction permits more accu- rate pixel-wise maps, avoiding errors due to respiratory deviations. An example of an image with and with- out motion correction is presented in Figure 4.

Clinical applications Potentially, T1 mapping can be used to assess any disease that affects the myocardium promoting diffuse fibrosis. However, because of its recent...
development, the technique has only been evaluated on a small number of patients although the clinical scenarios are varied.

The first clinical description of direct T1 mapping in pathological situations was done in patients with acute myocardial infarction [20]. While the authors did not use the described MOLLI sequence, they did note that pre-contrast infarct areas had an 18 ± 7% increase in T1 times compared to normal myocardium and that after contrast the same areas showed a 27 ± 4% reduction compared to non-infarcted areas (P < 0.05 for both). In chronic myocardial infarction, where LGE has proven to useful, these changes were also observed although differences were not as pronounced in the acute setting [21].

In amyloidosis, post-contrast T1 times were also detected to be shorter in the subendocardial regions compared to other myocardium areas [22]. The combination of both LGE identification and T1 times < 191 msec in the subendocardium at 4 minutes provided a 97% concordance in diagnosis of cardiac amyloidosis and T1 values significantly correlated to markers of amyloid load such as left ventricular mass, wall thickness, interstitial thickness and diastolic function.

In valve disease, an attempt to show changes in the overall group before or after contrast [23]. However, the authors did notice that differences were observed regionally in segments that demonstrated impaired wall motion in cine images. The small number of patients (n = 8) in the study might have affected the conclusions and further evaluation of similar data might yield other conclusions. More recent study showed that, using equivalence contrast CMR, diffuse fibrosis measured in aortic stenosis patients provided significant correlations to quantification on histology [24].

In heart failure, the use of T1 mapping has been more widely studied and directly correlated to histology evaluation [11]. In this paper, the authors evaluated patients with ischemic, idiopathic and restrictive cardiomyopathies showing that post-contrast T1 times at 1.5T were significantly shorter than controls even after exclusion of areas of LGE (429 ± 22 versus 564 ± 23 msec, P < 0.0001). We have investigated a similar group of patients on a 3T MAGNETOM Verio scanner and have found that both dilated and hypertrophic cardiomyopathy patients have lower post-contrast T1 times compared to controls, but non-infarcted areas from ischemic cardiomyopathy patients do not show significant differences (unpublished data).

Examples of a myocardial T1 map at 3T from a patient with dilated cardiomyopathy and suspected hypertrophic cardiomyopathy are seen in Figure 5 and 6 respectively.

Finally, in patients with both type 1 and 2 diabetes mellitus, T1 mapping using CMR was able to show that these patients may have increased interstitial fibrosis compared to controls as T1 times were significantly shorter (425 ± 72 msec versus 504 ± 34 msec, P < 0.001) and correlated to global longitudinal strain by echocardiography, demonstrating impaired myocardial systolic function.

Future directions

Certainly with the research of T1 mapping in different clinical scenarios the applicability of the method will increase substantially. In the mean time, more effort has been made to further standardize values across different patients and time points. As T1 time, especially after injection of contrast, depends on both physiologic and scan acquisitions, methods have been described to account for these factors, with normalization of T1 values [25]. More than that, standardization of normal values across a larger number of normal individuals is also necessary since most papers provide data on much reduced cohorts, mostly limited to single center data. In that regard, a large multicenter registry is already collecting data at 3T in patients from 20 to 80 years of age in Latin America (Fernandes IL et al. – www.clinicalatrialgs.org – NCT01230549). Besides that, other techniques are under development that might allow T1 measurement with larger coverage of the heart using 3D methods [26].

Nevertheless, with the current technical niches available there are already much more clinical applications to explore and certainly quantitative T1 mapping will become one of the key applications in CMR in the near future.

References

5A T1 mapping at 3T after contrast of a patient with (5A) dilated cardiomyopathy (T1 of 507 ms) in comparison to (5B) a control patient (T1 of 615 ms).

6A T1 mapping of a patient with (6A) suspected hypertrophic cardiomyopathy (T1 of 466 ms) in comparison to (6B) a control patient (with a T1 of 615 ms).


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Preliminary Experiences with Compressed Sensing Multi-Slice Cine Acquisitions for the Assessment of Left Ventricular Function: CV_sparse WIP

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Introduction

Left ventricular (LV) ejection fraction is one of the most important measures of outcome [1]. It allows to assess the effect of established or novel treatments [2], and it is crucial for decision making [3] e.g. to start [4] or stop [5] specific drug treatments or to implant devices [6]. CMR is generally accepted as the gold standard method to yield most accurate measures of LV ejection fraction and LV volumes. This capability and the additional value of CMR to characterize pathological myocardial tissue was the basis to assign a class I indication for patients with known or suspected heart failure to undergo CMR in the new Heart Failure Guidelines of the European Society of Cardiology [3].

The evaluation of LV volumes and LV ejection fraction are based on well-defined protocols [7] and it involves the acquisition of a stack of LV short axis cine images from which volumes are calculated by applying Simpson’s rule. These stacks are typically acquired in multiple breath-holds. Quality criteria [8] for these functional images are available and are implemented e.g. for the quality assessment within the European CMR registry which currently holds approximately 33,000 patients and connects 59 centers [9].

Recently, compressed sensing (CS) techniques emerged as a means to considerably accelerate data acquisition without compromising significantly image quality. CS has three requirements:

1) transform sparsity,
2) incoherence of undersampling artifacts, and
3) nonlinear reconstruction (for details, see below).

Based on these prerequisites, a CS approach for the acquisition of cardiac cine images was developed and tested [10]. In particular, the potential to acquire several slices covering the heart in different orientations within a single breath-hold would allow to apply model-based analysis tools which theoretically could improve the motion assessment at the base of the heart, where considerable through-plane motion on short-axis slices can introduce substantial errors in LV volume and LV ejection fraction calculations. Conversely, with a multi-breath hold approach, there are typically small differences in breath-hold positions which can introduce errors in volume and function calculations. The pulse sequence tested here allows for the acquisition of 7 cine slices within 14 heartbeats with an excellent temporal and spatial resolution.

Such a pulse sequence would also offer the advantage to obtain functional information in at least a single plane in patients unable to hold their breath for several heartbeats or in patients with frequent extrasystoles or atrial fibrillation. However, it should be mentioned that accurate quantitative measures of LV volumes and function cannot be obtained in highly arrhythmic hearts or in atrial fibrillation, as under such conditions volumes and ejection fraction change from beat to beat due to variable filling conditions. Nevertheless, rough estimates of LV volumes and function would still be desirable in arrhythmic patients.

In a group of healthy volunteers and patients with different LV pathologies, the novel single-breath-hold CS cine approach was compared with the standard multi-breath-hold cine technique with respect to measure LV volumes and LV ejection fraction. The CV_sparse work-in-progress (WIP) package implements sparse, incoherent sampling and iterative reconstruction for cardiac applications. This method in principle allows for high acceleration factors which enable triggered 20 real-time cine CMR while preserving high spatial and/or temporal resolution of conventional cine acquisitions. Compressed sensing methods exploit the potential of image compression during the acquisition of raw input data. Three components [10] are crucial for the concept of compressed sensing to work.

I. Sparsity: In order to guarantee compressibility of the input data, sparsity must be present in a specific transform domain. Sparsity can be computed e.g. by calculating differences between neighboring pixels or by calculating finite differences in angiograms which then detect primarily voxel contours which typically represent a few percent of the...
entire image data only. Furthermore, sparsity is not limited to the spatial domain: the acquisition of cine images of the heart can be highly sparsified in the temporal dimension.

II. Incoherent sampling: The aliasing artifacts due to k-space undersampling must be incoherent, i.e. noise-like, in that transform domain. Here, it is to mention that fully random k-space sampling is suboptimal incoherent sampling in space and time, the ICE program runs a non-orienting direction and/or the temporal offset is applied from frame-to-frame which results in an incoherent temporal jitter. Finally, a variable sampling density in k-space stabilizes the iterative reconstruction. To avoid eddy current effects for balanced steady-state free precession (bSSFP) acquisitions, pairing [13] can also be applied. Thus, the tested CV_spase sequence is characterized by sparse, incoherent sampling in space and time, non-linear iterative reconstruction integrating SENSE, and L1 wavelet regularization in the phase encoding direction and/or the temporal dimension. With regard to reconstruction, the ICE program runs a non-linear iterative reconstruction with k-space regularization in space and time specifically modified for compressed sensing. The algorithm derives from a parallel imaging type reconstruction which takes coil sensitivity maps into account, thus supporting predominantly high acceleration factors. For cine CMR, no additional reference scans are needed because – similar to TPAT – the coil sensitivity maps are generated by the analysis tool. Of note, this 4D analysis tool automatically tracks the 3-dimensional motion of the mitral annulus throughout the cardiac cycle which allows for an accurate volume calculation particularly at the base of the heart.

III. Reconstruction: A non-linear iterative optimization corrects for sub-sampling artifacts during the process of image reconstruction yielding to a better solution with a sparse representation in a specific transform domain and which is consistent with the input data. Such compressed sensing techniques can also be combined with parallel imaging techniques [11].

WIP CV_sparse Sequence

The current CV_spase sequence [12] realizes incoherent sampling by initially distributing the readouts pseudo-randomly on the Cartesian grid in k-space. In addition, for cine-CMR imaging, a pseudo-random offset is applied from frame-to-frame which results in an incoherent temporal jitter. Finally, a variable sampling density in k-space stabilizes the iterative reconstruction. To avoid eddy current effects for balanced steady-state free precession (bSSFP) acquisitions, pairing [13] can also be applied. Thus, the tested CV_spase sequence is characterized by sparse, incoherent sampling in space and time, non-linear iterative reconstruction integrating SENSE, and L1 wavelet regularization in the phase encoding direction and/or the temporal dimension. With regard to reconstruction, the ICE program runs a non-linear iterative reconstruction with k-space regularization in space and time specifically modified for compressed sensing. The algorithm derives from a parallel imaging type reconstruction which takes coil sensitivity maps into account, thus supporting predominantly high acceleration factors. For cine CMR, no additional reference scans are needed because – similar to TPAT – the coil sensitivity maps are calculated from the temporal average of the input data in a central region of k-space consisting of not more than 48 reference lines. The extensive calculations for image reconstruction typically running 80 iterations are performed online on all CPUs on the MARS computer in parallel, in order to reduce reconstruction times.

Volunteer and Patient studies

In order to obtain insight into the image quality of single-breath-hold multi-slice cine CMR images acquired with the compressed sensing (CS) approach, we studied a group of healthy volunteers and a patient group with different pathologies of the left ventricle. In addition to the evaluation of image quality, the robustness and the precision of the CS approach for LV volumes and LV ejection fraction was also assessed in comparison with a standard high-resolution cine CMR approach. All CMR examinations were performed on a 1.5 T MAGNETOM Aera (Siemens Healthcare, Erlangen, Germany). The imaging protocol consisted of a set of cardiac localizers followed by the acquisition of a stack of conventional short-axis SSFP cine images covering the entire LV with a spatial and temporal resolution of 1.2 x 1.6 mm², and approximately 40 ms, respectively (slice thickness: 8 mm; gap between slices: 2 mm). LV 2-chamber, 3-chamber, and 4-chamber long-axis acquisitions were obtained for image quality assessment but were not used for LV volume quantifications. As a next step, to test the new CS-based technique, slice orientations were planned to cover the LV with 4 short-axis slices distributed evenly over the LV long axis complemented by 3 long-axis slices (i.e., 2-chamber, 3-chamber, and 4-chamber slice) (Fig. 1). These 7 slices were then acquired in a single breath-hold maneuver lasting 14 heart beats (i.e. 2 heart beats per slice) resulting in an acceleration factor of 11.0 with a temporal and spatial resolution of 30 ms and 1.5 x 1.5 mm², respectively (slice thickness: 6 mm). As the reconstruction algorithm is susceptible to aliasing in the phase-encoding direction, the 7 slices were first acquired with a non-cine acquisition to check for correct phase-encoding directions and, if needed, to adjust the field-of-view to avoid fold-over artifacts. After confirmation of correct imaging parameters, the 7 slice single-breath-hold cine CS-acquisition was performed. In order to obtain a reference for the LV volume measurement, a phase-contract flow measurement in the ascending aorta was performed to be compared with the LV stroke volumes calculated from the standard and CS cine data.

The conventional stack of cine SSFP images was analyzed by the Argus software (Siemens Argus 4D Ventricular Function, Fig. 2A). The CS cine data were analyzed by the 4D-Argus software (Siemens Argus, Fig. 2B). Such software is based on an LV model and, with relatively few operator interactions, the contours for the LV endocardium and epicardium are generated by the analysis tool. Of note, this 4D analysis tool automatically tracks the 3-dimensional motion of the mitral annulus throughout the cardiac cycle which allows for an accurate volume calculation particularly at the base of the heart.
olution may occur in relation to the patient’s anatomy. In addition, the sparsity in the temporal domain may be limited in anatomical regions of very high flow, and therefore, in some acquisitions, flow-related artifacts occurred in the phase-encoding direction during systole (Figs. 6B, C). Also, in its current version, the sequence is prospective, thus it does not cover the very last phases of the cardiac cycle and the reconstruction times for the CS images lasted several minutes precluding an immediate assessment of the image data quality or using this image information to plan next steps of a CMR examination.

Performance of the single-breath-hold CS approach in comparison with the standard multi-breath-hold cine approach

From a quantitative point-of-view, the accurate and reliable measurement of LV volumes and function is crucial as many therapeutic decisions directly depend on these measures [3–6]. In this current relatively small study group, LV end-diastolic and end-systolic volumes measured by the single-breath-hold CS approach were comparable with those calculated from the standard multi-breathhold cine SSFP approach. LVEDV and LVESV differed by 10 ml ± 17 ml and 2 ml ± 12 ml, respectively. Most importantly, LV ejection fraction differed by only 1.3 ± 4.7% (50.6% vs 49.3%) for multi-breath-hold and single-breathhold, respectively, p = 0.77, regression: r = 0.96, p < 0.0001; y = 0.96x + 0.8 ml). Thus, it can be concluded that the single-breath-hold CS approach could potentially replace the multi-breath-hold standard technique for the assessment of LV volumes and systolic function.

What about the accuracy of the novel single-breath-hold CS technique?

To assess the accuracy of the LV volume measurements, LV stroke volume was compared with the LV output measured in the ascending aorta with phase-contrast MR. As the flow measurements were performed distally to the coronary arteries, flow in the coronary arteries was estimated as the LV mass multiplied by 0.8 ml/min/g. Excellent agreement was found with a mean of 86.8 ml/beat for the aortic flow measurement and 91.9 ml/beat for the LV measurements derived from the single-breath-hold CS data (r = 0.93, p < 0.0001). By Bland-Altman analysis, the stroke volume approach overestimated by 5.2 ml/beat versus the reference flow measurement. For the conventional stroke volume measurements, this difference was 15.6 ml/beat (linear regression analysis vs aortic flow: r = 0.69, p < 0.01). More importantly, the CS LV stroke data were not only more precise with a smaller mean difference, the variability of the CS data vs the reference flow data was less with a standard deviation as low as 6.8 ml/beat vs 12.9 ml/beat for the standard multi-breath-hold approach (Fig. 7). Several explanations may apply for the higher accuracy of the single-breath-hold multi-slice CS approach in comparison to the conventional multi-breath-hold approach:

1) With the single-breath-hold approach, all acquired slices are correctly co-registered, i.e. they are correctly aligned in space, a prerequisite for the 4D-analysis tool to work properly.

2) This 4D-analysis tool allows for an accurate tracking of the mitral valve plane motion during the cardiac cycle as shown in Figure 5, which is important as the cross-sectional area of the heart at its base is large and thus, inaccurate slice positioning at the base of
axis slices typically translate in the heart with conventional short-axis slices, with the LV blood pool contour in the end-systolic phase, are excluded from the LV walls [15]. From Figures 8A, 8B, we observed a systematic overestimation of LV end-diastolic volume. Small trabeculations (yellow contours in 8A) are included into the LV blood pool resulting in a small underestimation of end-systolic volume, and thus, LV stroke volume. This explanation is likely as Van Rossum et al. demonstrated a slight underestimation of the LV mass when calculated on end-diastolic phases versus end-systolic phases, as trabeculations in end-diastole are typically excluded from the LV walls [15]. In summary, this novel very fast multi-slice cine technique allows to cover the entire LV with high temporal and spatial resolution within a single breath-hold. The image quality based on these preliminary results appears adequate to yield highly accurate measures of LV volumes, LV stroke volume, LV mass, and LV ejection fraction.

The Cardiac MR Center of the University Hospital Lausanne

The Cardiac Magnetic Resonance Center (CMRC) of the University Hospital of Lausanne (Centre Hospitalier Universitaire Vaudois, CHUV) was established in 2009. The CMR center is dedicated to high-quality clinical work-up of cardiac patients, to deliver state-of-the-art training in CMR to cardiologists and radiologists, and to pursue research.

In the CMR center education is provided for two specialties while focusing on one organ system. Traditionally, radiologists have focused on using one technique for different organs, while cardiologists have concentrated on one organ and perhaps one technique. Now in the CMR center the focus is put on a combination of specialists with different background on one organ. Research at the CMR center is devoted to four major areas: the study of 1) cardiac function and tissue characterization, specifically to better understand diastolic dysfunction, 2) the development of MR-compatible cardiac devices such as pacemakers and ICDs; 3) the utilization of hyperpolarized 13C-carbon contrast media to investigate metabolism in the heart, and 4) the development of 1F-fluorine-based CMR techniques to detect inflammation and to label and track cells non-invasively.

For the latter two topics, the CMR center established tight collaborations with the Center for Biomedical Imaging (CIBM), a network around Lake Geneva that includes the École Polytechnique Fédérale de Lausanne (EPFL), and the universities and university hospitals of Lausanne and Geneva. In particular, strong collaborative links are in place with the CMR team of Prof. Matthias Stubler, a part of the CIBM and located at the University Hospital Lausanne and with Prof. A. Comment, with whom we perform the studies on real-time metabolism based on the 13C-carbon hyperpolarization (DNP) technique. In addition, collaborative studies are ongoing with the Heart Failure and Cardiac Transplantation Unit led by Prof. R. Hullin (detection of graft rejection by tissue characterization) and the Oncology Department led by Prof. Coukos (T cell tracking by 19F-MRI in collaboration with Prof. Stuber, R. van Heeswijk, CIBM, and Prof. O. Michielin, Oncology). This structure allows for a direct interdisciplinary interaction between physicists, engineers, and basic scientists on a daily basis with the aim to enable innovative research and fast translation of these techniques from bench to bedside.

The CMRC is also the center of competence for the quality assessment of the European CMR registry which holds currently approximately 33,000 patient studies acquired in 59 centers across Europe. The members of the CMRC team are: Prof. J. Schwittek (director of the center), PD Dr. X. Jeannenay, Dr. D. Locca, MER, Dr. P. Monney, Dr. T. Rutz, Dr. C. Sierro, and Dr. S. Koestner (cardiologists, staff members).

An excellent correlation is obtained for the LV stroke volume calculated from the compressed sensing data with the flow volume in the aorta measured by phase-contrast technique. Variability of the conventional LV stroke volume data appears higher than for the compressed sensing data.
Acknowledgements

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Accelerated Segmented Cine TrueFISP of the Heart on a 1.5T MAGNETOM Aera Using k-t-sparse SENSE

Introduction

Cine MRI of the heart is widely regarded as the gold standard for assessment of left ventricular volume and myocardial mass and is increasingly utilized for assessment of cardiac anatomy and pathology as part of clinical routine. Conventional cine imaging approaches typically require 1 slice per breath-hold, resulting in lengthy protocols for complete cardiac coverage. Parallel imaging allows some shortening of the acquisition time, such that 2–3 slices can be acquired in a single breath-hold. In cardiac cine imaging artifacts become more prevalent with increasing acceleration factor. This negates the diagnostic utility of the images and may reduce accuracy of quantitative measurements. However, regularized iterative reconstruction techniques can be used to considerably improve the images obtained from highly undersampled k-space data. This technique* takes advantage of the de-noising characteristics of Wavelet regularization and promises to very effectively suppress subsampling artifacts. This may allow for high acceleration factors to be used, while diagnostic image quality is preserved.

The purpose of this study was to compare segmented cine TrueFISP images from a group of volunteers and patients using three acceleration approaches: iPAT factor 4 with conventional reconstruction; T-PAT factor 4 with conventional reconstruction; and T-PAT factor 4 with iterative k-t-sparse SENSE reconstruction.

Technique

Cardiac MRI seems to be particularly well suited to benefit from a group of novel image reconstruction methods known as compressed sensing [2] which promise to significantly speed up data acquisition. Compressed sensing methods were introduced to MR imaging [3, 4] ago and have since been successfully combined with parallel imaging [5, 6]. Such methods try to utilize the

Table 1: MRI conventional and iterative imaging parameters

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Conventional iPAT 2</th>
<th>Conventional T-PAT 4</th>
<th>Iterative T-PAT 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Iterative recon</td>
<td>No</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Parallel imaging (PAT2 (GRAPPA))</td>
<td>TPA2</td>
<td>TPA2</td>
<td>TPA2</td>
</tr>
<tr>
<td>TR/TE (ms)</td>
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<td>3.2/1.6</td>
<td>3.2/1.6</td>
</tr>
<tr>
<td>Flip angle (degrees)</td>
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<td>70</td>
<td>70</td>
</tr>
<tr>
<td>Pixel size (mm)²</td>
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<td>1.9 x 1.9</td>
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</tr>
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<td>8</td>
</tr>
<tr>
<td>Temp. res. (msec)</td>
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<td>38</td>
<td>38</td>
</tr>
<tr>
<td>Acq. time (sec)</td>
<td>7</td>
<td>3.2</td>
<td>3.2</td>
</tr>
</tbody>
</table>

*Work in progress. The product is still under development and not commercially available yet. Its future availability cannot be ensured.
As outlined by Liu et al. in [1], the image reconstruction can be formulated as an unconstrained optimization problem. In the current implementation, this optimization is solved using a Nesterov-type algorithm [7]. The L1-regularization with a redundant Haar transform is efficiently solved using a Dykstra-type algorithm [8]. This allowed a smooth integration into the current MAGNETOM platform and, therefore, facilitates a broad clinical evaluation.

Materials and methods

Nine healthy human volunteers (57.4 male/56.7 female) and 20 patients (54.4 male/46.0 female) with suspected cardiac disease were scanned on a 1.5T MAGNETOM Aera system under an approved institutional review board protocol. All nine volunteers and 16 patients were imaged using segmented cine TrueFISP sequences with conventional GRAPPA factor 2 acceleration (conventional iPAT 2), T-PAT factor 4 acceleration (conventional T-PAT 4), and T-PAT factor 4 acceleration with iterative k-space SENSE reconstruction (iterative T-PAT 4). The remaining 4 patients were scanned using only conventional iPAT 2 and iterative T-PAT 4 techniques. Note that the iterative technique is fully integrated into the standard reconstruction environment.

The imaging parameters for each imaging sequence are provided in Table 1. All three sequences were run in 3 chamber and 4 chamber views, as well as a stack of short axis slices. Quantitative analysis was performed on all volunteer data sets at a syngo MultiModality Workplace (Leonardo) using Argus post-processing software (Siemens Healthcare, Erlangen, Germany) by an experienced cardiovascular MRI technician. Ejection fraction, end-diastolic volume, end-systolic volume, stroke volume, cardiac output, and myocardial mass were calculated. In all volunteers and patients, blinded qualitative scoring was performed by a radiologist using a 5 point Likert scale to assess overall image quality (1 – non diagnostic, 2 – poor, 3 – fair, 4 – good, 5 – excellent). Images were also scored for artifact and noise (1 – severe, 2 – moderate, 3 – mild, 4 – trace, 5 – none).

All continuous variables were compared between groups using an unpaired t-test, while ordinal qualitative variables were compared using a Wilcoxon signed-rank test.

Results

All images were acquired successfully and image quality was of diagnostic quality in all cases. The average scan time per slice for conventional iPAT 2, conventional T-PAT 4 and iterative T-PAT 4 were for patients 7.7 ± 1.5 sec, 5.6 ± 1.5 sec and 2.9 ± 1.5 sec and for the volunteers 9.8 ± 1.5 sec, 3.2 ± 1.5 sec and 3.0 ± 1.5 sec, respectively. The results in scan time are illustrated in Figure 1. In both patients and volunteers, conventional iPAT 2 were significantly longer than both conventional T-PAT 4 and iterative T-PAT 4 techniques (p < 0.001 for each group).

Discussion

This study compares a novel accelerated segmented cine TrueFISP technique to conventional iPAT 2 cine TrueFISP and T-PAT 4 cine TrueFISP in a cohort of normal subjects and patients. The iterative reconstruction technique provided comparable measurements of ejection fraction to the clinical gold standard (conventional iPAT 2). The accelerated segmented cine TrueFISP with T-PAT 4, which was used as comparison technique, produced slightly lower EF values compared to the other techniques, although this was not found to be statistically significant. The iterative reconstruction produced comparable image quality, noise and artifact scores to the conventional reconstruction using iPAT 2. The conventional T-PAT 4 technique had lower image quality and higher noise scores compared to the other two techniques. The iterative T-PAT 4 segmented cine technique allows for greater than 50% reduction in acquisition time for comparable image quality and spatial resolution as the clinically used iPAT 2 cine TrueFISP technique. This iterative technique could be extended to permit complete heart coverage in a single breath-hold thus greatly simplifying and shortening routine clinical cardiac MRI protocols, which has been one of the biggest obstacles to wide acceptance of cardiac MRI. With a shorter cine acquisition, additional advanced imaging techniques, such as perfusion and flow, can be more readily added to patient scans within a reasonable protocol length.
There are currently some limitations to the technique. Firstly, the use of SENSE implies that aliasing artifacts can occur if the field-of-view is smaller than the subject, which is sometimes difficult to avoid in the short axis orientation. But a solution to this is promised to be part of a future release of the current prototype. Secondly, the image reconstruction times of the current implementation seem to be prohibitive for routine clinical use. However, we anticipate future algorithmic improvements with increased computational power to reduce the reconstruction time to clinically acceptable values.

Of course, iterative reconstruction techniques are not just limited to cine imaging of the heart. Future work may see this technique applied to time intense techniques such as 4D flow phase contrast MRI and 3D coronary MR angiography, making them more clinically applicable. Furthermore, higher acceleration rates might be achieved by using an incoherent sampling pattern [9].

With sufficiently high acceleration, the technique can also be used effectively for real time cine cardiac imaging in patients with breath-holding difficulties or arrhythmia. Figure 6 shows that real-time acquisition with T-PAT 6 and k-t iterative reconstruction still results in excellent image quality.

In conclusion, cine TrueFISP of the heart with inline k-t sparse iterative reconstruction is a promising technique for obtaining high quality cine images at a fraction of the scan time compared to conventional techniques.

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Combined ¹⁸F-FDG PET and MRI Evaluation of a case of Hypertrophic Cardiomyopathy Using Simultaneous MR-PET

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Introduction

Hypertrophic cardiomyopathy (HCM) is a common condition causing left ventricular outflow obstruction, as well as cardiac arrhythmias. Cardiac MRI is a key modality for evaluation of HCM. Apart from estimating left ventricular (LV) wall thickness, LV function and aortic flow, MRI is capable of estimating the late gadolinium enhancement in affected myocardium, presumably related to microvascular disease [2]. ¹⁸F-FDG PET has been sporadically studied in HCM, mostly for evaluation of the metabolic status of the hypertrophic myocardial segment, especially after interventions such as transcatheter ablation of septal hypertrophy (TASH) [3] or to demonstrate partial myocardial fibrosis [4]. This clinical example illustrates the value of integrated simultaneous ¹⁸F-FDG PET and MRI acquisition performed on the Biograph mMR system.

Patient history

A 25-year-old man presented to the cardiology department with incidental ECG abnormality after fractures to his left 2nd and 4th fingers. Although he had not consulted a doctor, he had been suffering from mild dyspnea with chest discomfort at rest and exacerbation at exercise since May 2012. Echocardiography revealed non-obstructive hypertrophic cardiomyopathy (Maron III) with trivial MR. The patient was referred for a simultaneous MR-PET study for ¹⁸F-FDG PET and cardiac MRI with Gadolinium (Gd) contrast for evaluation of the severe symptoms of this patient.

Discussion

The late Gd enhancement within the hypertrophic septum along with the non-uniform glucose metabolism demonstrated by the patchy ¹⁸F-FDG uptake within the hypertrophic septum exactly corresponding to the area of Gd enhancement reflect myocardial fibrosis within the asymmetric septal hypertrophy. Myocardial fibrosis and the presence of late Gd enhancement on MRI has been shown to be associated with increased risk of cardiac arrhythmia [1] as evident from the symptoms of this patient.

Simultaneous MR-PET acquisition provides combined acquisition of both modalities, thereby ensuring an accurate fusion between morphological and functional images due to the comprehensive cardiac MRI sequences were acquired including MR perfusion after Gd contrast infusion, as well as post contrast late Gd enhancement studies. Static ¹⁸F-FDG PET was acquired simultaneously during the MRI acquisition.
Cardiac MRI at 3T: An Indian Experience of 80 Cases

Cardiac MRI is the main investigation modality for a wide range of clinical applications and has emerged as a virtual ‘one-stop-shop’ for imaging conditions such as Cardiomyopathies. CMR has added uniquely to the methods for non-invasive assessment of myocardial viability by a combination of cine imaging and delayed hyper-enhancement. CMR provides excellent depiction of pericardium in conditions such as pericarditis, pericardial effusions, and masses. It provides optimal assessment of the location, functional characteristics, and soft tissue features of cardiac tumors, allowing accurate differentiation of benign and malignant lesions. MRI is ideally suited to serve as the primary imaging modality in patients with congenital heart disease due to its non-invasive and biologically harmless nature, and its ability to provide accurate anatomical and functional information. Several investigators have confirmed the SNR advantages of CMR at 3T. These indicate an overall quantitative improvement in SNR and CNR, thus improving imaging capabilities.

Dr. Bidarkar and colleagues (Departments of Radiology and Cardiology, Jupiter Hospital, Thane, Maharashtra, India) illustrate 80 cases of cardiac MRI imaged between January 2012 and August 2013 on a 3T MAGNETOM Vero.

Read the comprehensive article
www.siemens.com/magnetom-world

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Molecular MRI in London Ontario

The Lawson Health Research Institute (the “Lawson”) is located within St. Joseph’s Healthcare in London Ontario, and is affiliated with Western University. This group received the first Biograph mMR in Canada in March 2012. With strong existing research groups in MRI and nuclear medicine this group is ideally positioned to drive research and innovation using this platform. In advance of the installation of the hybrid magnetic resonance, molecular magnetic resonance, researchers at the Lawson had been developing novel medical and radiochemistry facilities, and novel tracer development methods. In addition to the molecular MR site this has a PET-CT, a SPECT-CT, and an Inveon small animal PET as well as two 1.5T MAGNETOM Aera systems, a 16.5MeV medical cyclotron and radiochemistry facilities.

Myocardial tissue imaging

Hybrid imaging platforms incorporating PET have become available for cardiovascular imaging applications over the past decade. These platforms have been primarily aimed at providing superior tissue attenuation correction of the emitted photon signal and to provide spatial anatomic registration for the localization of abnormal tracer signal. While this has resulted in substantial improvements in the clinical performance of cardiac PET, the exploitation of complementary imaging data has yet to be fully realized.

The recent availability of hybrid platforms allows for an expansive range of PET applications to be explored. For example the capacity of cardiovascular MRI to provide complementary 2D and 3D morphological data with excellent soft tissue contrast and high temporal resolution is of benefit for anatomic registration and novel motion correction algorithms. However, its incremental capacity to provide exquisitely tissue characterization through intrinsic tissue contrast and altered kinetics of exogenous paramagnetic contrast is of particular interest in the context of the PET imaging environment.

Clinical adoption of myocardial tissue imaging is expanding in response to mounting evidence that the ‘health’ of myocardium strongly modulates benefit from heart failure therapies (pharmacologic, surgical and device-based) and is predictive of future arrhythmic events among patients with ischemic and non-ischemic cardiomyopathy [1-5]. To date, this literature has focused on isolated and disparate markers of tissue health using both PET and MRI. However, within this brief report we discuss several synergies of these platforms that hold promise towards a new era of hybrid imaging for the optimal performance of myocardial tissue imaging.

Molecular MRI in the setting of acute ischemic injury

Among those surviving acute myocardial infarction (AMI), appropriate myocardial healing is believed to be reliant upon a highly choreographed process of early inflammatory cell invasion, collagen degradation, debris removal by activated macrophages, and myofibroblast proliferation with reconstitution of a new collagen matrix. While such findings can be characterized histologically, our capacity to quantify markers of the inflammatory process in vivo, and evaluate influences of its modulation on the remodeling process has been limited.

We have started examining this process in a canine infarct model using the Biograph mMR 3T-PET platform using simultaneous 3D LGE (T1-FDG imaging, the latter imaged following normal myocardial glucose metabolism using intravenous heparin and lipid infusion. In these experiments we have focused on evaluating the influence of microvascular obstruction (MO) on mitigating appropriate inflammatory cell recruitment to the infarct core – a postulated mechanism of how MO may adversely impact on left ventricular remodeling post infarct. Figure 2 illustrates how the region of MO can be elegantly visualized using both LGE imaging and T1 mapping CMR techniques. 18F-FDG imaging shows intense inflammatory activity within the perfused infarct rim, however a marked reduction in activity is seen in regions of MO. This imaging may therefore provide novel insights towards mechanisms by which MO contributes to adverse outcomes following AMI, and offers a new tool to evaluate therapies aimed at modulation of this pathway.

Molecular MRI in the setting of non-ischemic (inflammatory) injury

Both PET and CMR have been investigated for their diagnostic accuracy in the setting of suspected inflammation – particularly among patients with known pulmonary Sarcoid. While their respective diagnostic performance has been compared in the past, this remains inappropriate, as the information gathered and interpreted from each technique could not be more unique.

PET imaging (typically performed using 18F-FDG following prolonged fasting, fatty meal consumption and intravenous heparin to suppress normal myocardial glucose utilization) exploits the hypermetabolic signal of activated inflammatory cells (i.e. macrophages) and therefore indicates disease ‘activity’ among patients with active cardiac Sarcoid. In contrast, LGE imaging indicates regions of mature granulomatous fibrosis among patients with prior or current cardiac Sarcoid. Therefore, these two commonly employed diagnostic techniques provide complementary but unique information.
Figure 3 illustrates the capacity of hybrid MR and PET to spatially register these techniques using simultaneously acquired data, and potentially improve diagnostic accuracy while expanding our understanding of disease pathophysiology. In this case of a 72-year-old female presenting with heart failure and non-sustained ventricular tachycardia we can identify a leading edge of inflammation (intense FDG uptake) with a trailing edge of irreversible injury or ‘scar’ (indicated by hyperenhancement on LGE imaging) at the sub-epicardial zone. This approach ushers in a new marriage of PET-based metabolic imaging and LGE-based scar imaging providing a more robust platform for the prediction of improved outcomes following coronary revascularization.

In figure 4 we see a 42-year-old male referred for viability imaging prior to coronary artery bypass surgery late following myocardial infarction. Cine imaging shows a large region of thinned and akinetik myocardium in the distribution of the left anterior descending artery, this territory demonstrating varying degrees of transmural scar following gadolinium administration. Simultaneous MR and PET with °FDG imaging shows metabolic activity within non-scared regions surrounding the infarct zone and normal metabolic tracer activity within remote myocardium.

Making is for the assessment of tissue viability in chronic ischemic cardiomyopathy. Evidence supports that both the regional reduction of FDG signal and LGE are strongly predictive of absence of functional recovery following coronary revascularization. The performance of these studies are therefore commonly considered to be mutually exclusive, with absence of FDG signal being the sine qua non of myocardial scar, and the lack of scar by LGE imaging being equivalent to tissue health. However, it is recognized that the spatial resolution and signal-to-noise of LGE imaging is superior to FDG for the detection of subendocardial scar, and also for the characterization of tissue viability among those with marked thinning of the ventricular wall [8]. Conversely, FDG-PET based metabolic abnormalities can be documented within tissue that fails to demonstrate myocardial scar on LGE MRI, lack of FDG uptake being predictive of absence of functional recovery. Accordingly, the marriage of PET-based metabolic imaging and LGE-based scar imaging may provide a more robust platform for the prediction of improved outcomes following coronary revascularization.

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