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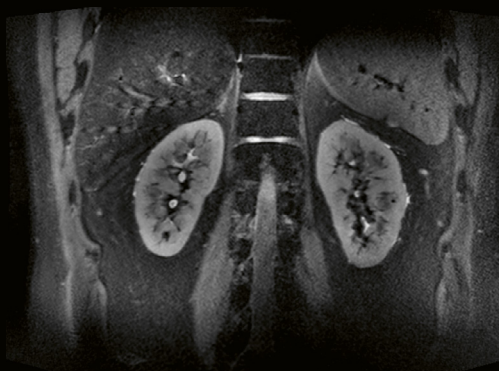
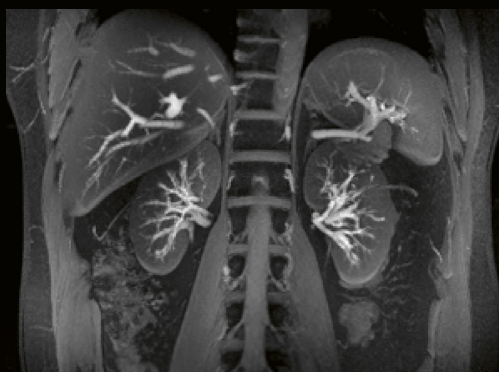
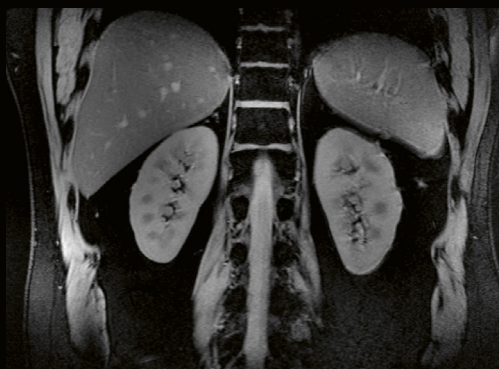
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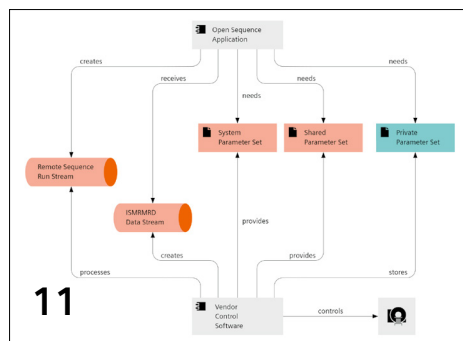
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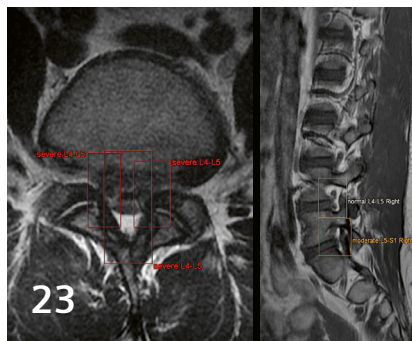
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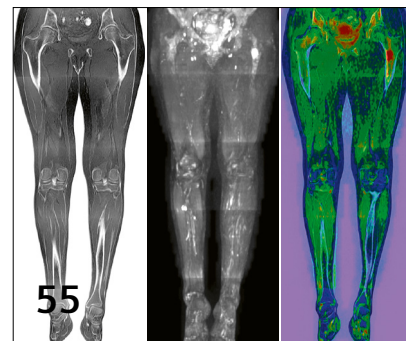




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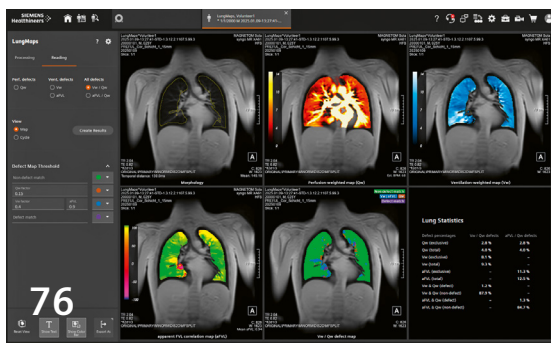
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¹ Work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

² The statements for energy and cost savings described herein are based on results that were achieved in settings possible due to certain standard and optional features available on MAGNETOM systems. Because there is no "typical" MRI suite or installation and many variables exist there can be no guarantee that customers will achieve the same results.

³ Abdominal imaging at ultra-high-field MRI is a research application and is not released for clinical use. It is not for sale in the U.S. and in other countries and its future availability cannot be ensured.

⁴ MR scanning has not been established as safe for imaging fetuses and infants less than two years of age. The responsible physician must evaluate the benefits of the MR examination compared to those of other imaging procedures.

⁵ MAGNETOM Iseult is an investigational device only and is not intended for sale or clinical use.



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Charting New Waters: The Evolution of Body MRI from Technical Marvel to Clinical Cornerstone

The past decade has witnessed a remarkable transformation in body MRI, with the technique evolving from a specialized problem-solving tool to an indispensable clinical workhorse. This evolution represents not a reduction in capability but rather a culmination of technological refinement that is making sophisticated imaging more accessible, efficient, and clinically impactful.

The technical renaissance in body MRI

In the early 2010s, body MRI was often considered time-consuming, technically challenging, and reserved for specific diagnostic dilemmas. The improvement of parallel imaging techniques such as CAIPIRINHA has consistently allowed for acceleration factors of 4 (2×2) in both frequency and phase-encoding directions [1], enabling faster acquisition with reasonable spatial resolution in abdominal imaging [2, 3].

By the mid-2010s, compressed sensing was finally implemented in commercial scanners, allowing single-breath-hold 3D MR cholangiopancreatography – one of the longest sequences in body MRI [4, 5] – and faster dynamic T1-weighted images for stable body imaging [6, 7]. Furthermore, technical developments offered clinically available free-breathing T1-weighted imaging using GRASP and XD-VIBE sequences for patients who cannot

hold their breath sufficiently, and for pediatric¹ populations [8–10]. These techniques contribute to accelerating scan speed, improving workflows, and reducing recall rates. In addition to radial sequences, ultra-short-TE² and zero-TE techniques now provide acceptable image quality for lung MRI in fibrotic disease and lung nodule detection [11, 12], as shown in the article by Serai and Rapp in this issue of MAGNETOM Flash [13].

Now, in the 2020s, deep learning (DL) reconstruction is being used for 2D and 3D sequences in body imaging. DL reconstruction, including Deep Resolve, produces better image quality in shorter acquisition times for T2-weighted imaging, diffusion-weighted imaging (DWI), and dynamic T1-weighted sequences [14–17]. The integration of SAMER retrospective motion correction with DL reconstruction² in brain imaging, described by Polak and colleagues in this issue [18], addresses one of the most persistent challenges in MRI: patient motion. It exemplifies how artificial intelligence has transformed our ability to obtain diagnostic images in a fraction of the time previously required.

As a result, the landscape has dramatically changed. The convergence of hardware innovations – particularly in gradient performance, coil technology, and sequence design – has eliminated many limitations. The articles in this issue highlight several of these advancements.

Navigating microstructures with innovative technology

Exploring tissue and cells noninvasively and in vivo is always a fascinating topic in this field, and DWI is a tailored sequence for exploring microstructures in the body. Beyond its clinical value for detecting or characterizing malignancies, DWI serves as a useful probe to navigate tissue structure: Non-Gaussian models (intravoxel incoherent motion, diffusion kurtosis imaging, stretched model, time-dependent DWI, and oscillating gradient DWI) have been reported to quantitatively assess pathological tissue in abdominal organs [19–23].

However, these non-Gaussian models remain largely in the research domain due to limitations including long scan times, limited signal-to-noise ratio (SNR), and reproducibility challenges. Most clinical scanners offer gradient systems with maximum amplitudes of 40–80 mT/m, which lead to relatively long TE and decreased SNR. From the Human Connectome Project, we learned how scanners with high gradient performance (300–500 mT/m) revolutionized microstructural imaging [24]. This high-performance technology has been successfully translated into a clinical platform in the form of MAGNETOM Cima.X, which provides 200 mT/m of gradient strength.

The promise of powerful gradients shown in brain imaging is also evident in prostate imaging. In this issue of MAGNETOM Flash, Molendowska and colleagues highlight how stronger gradient systems (200–500 mT/m compared to clinical 40–80 mT/m) enable shorter diffusion encoding times or allow higher b-values, improving the trade-off between SNR and TE [25]. Together with tailored gradient encoding and more efficient image readout strategies such as spiral readouts and advanced reconstruction techniques, SNR and contrast-to-noise ratio are improved in advanced DWI models. Most encouragingly, these technical advances translate into clinically significant changes in prostate cancer trials [26].

Another ongoing development is the application of ultra-high-field MRI for body imaging³. Ultra-high-field MRI has proven its potential for visualizing anatomical substructures in neuroimaging with higher SNR and spatial resolution [27, 28], as highlighted by Kurz and colleagues in their article using the second-generation clinical 7T whole-body MRI scanner MAGNETOM Terra.X [29]. Furthermore, increased SNR and high spectral resolution allow non-proton nuclei imaging such as ²³Na and ³¹P [28]. Is 7T applicable to body imaging? Attempts are scarce due to concerns about B1+ inhomogeneity and SAR. Metzger and colleagues explain how they tackled obstacles in body imaging at 7T with technical advances in RF coils and dynamic parallel transmission [30]. While still in the early stages, these approaches provide more specific microstructural information that may improve diagnosis, staging,

and treatment monitoring non-invasively, potentially contributing to personalized medicine.

From dedicated MRI to versatile whole-body imaging

MRI is often used to evaluate specific single organs with dedicated sequences, while CT scanning is regarded as the optimal modality in oncologic patients for staging, follow-up, and surveillance in high-risk groups. However, technical advances now allow larger scan coverage with body MRI. This progress has been particularly notable in oncological applications. Whole-body MRI (WB-MRI) has proven its role in oncologic patients, especially in those whose cancer has bone marrow involvement (such as myeloma, lymphoma, and bone metastases), and in pediatric¹ populations [31–35].

Standardized protocols and reporting systems have been suggested, with MY-RADS for multiple myeloma [36], ONCO-RADS for cancer screening [37], and MET-RADS-P for advanced prostate cancer [38]. However, current standard WB-MRI protocols have limited capability to visualize mineralized bone. The work by McGuire and colleagues shows how proton-density-weighted VIBE sequences with Deep Resolve Boost² have become a viable solution for visualizing mineralized bone by creating CT-like images in a reasonable timeframe thanks to DL [39]. Further, DL-based body segmentation offers additional information for patient prognoses based on WB-MRI [40]. No longer constrained by the traditional limitations of examination time and patient tolerance, WB-MRI now offers a more complete solution for many oncologic patients.

The PET/MRI synergy

The convergence of functional and anatomical imaging through PET/MRI represents another frontier in oncological imaging. As detailed in Schmidt's historical perspective on the development of integrated PET/MRI scanners, this technology has evolved from conceptual discussions in the late 1990s to a clinical reality with Biograph mMR (Siemens Healthineers, Erlangen, Germany) in 2010–2011 [41]. The system uniquely combines the anatomical resolution of MRI with the sensitivity of PET, enabling simultaneous and isocentric imaging.

This synergistic approach has proven particularly valuable in oncological imaging, where the correlation between functional activity and anatomical abnormalities is critical for accurate staging and tumor response evaluation [42–44]. Compared with PET/CT, PET/MRI has demonstrated better performance for liver and brain lesions where background uptake is high [45]. Additionally, with new tracers such as DOTATOC and DOTATATE for

neuroendocrine tumors [46], PSMA for prostate cancer [47], and FAPI for breast and pancreatic cancers [48], the clinical indications for PET/MRI continue to broaden.

Now, 14 years after the introduction of Biograph mMR, the technology continues to advance with the upcoming BIOGRAPH One² system, which integrates key technologies from both PET and MRI systems to address system complexity and improve throughput. This integration brings PET/MRI closer to its original goal: an all-in-one modality combining submillimeter anatomical details with functional imaging. With expanded clinical applications for theranostics [49, 50], PET/MRI continues to shape the era of personalized care.

The new kids on the block

MRI has long since established itself as a powerful problem-solving tool in radiology, but its emerging role in cancer surveillance represents an exciting paradigm shift in oncology. While traditional screening methods such as mammography for breast cancer, PSA for prostate cancer, and ultrasound for hepatocellular carcinoma have been standard practice for decades, MRI has consistently demonstrated superior sensitivity and specificity across multiple cancer types, enabling early detection of small tumors [51–53]. The non-invasive nature of MRI, coupled with its exceptional soft tissue contrast resolution, makes it uniquely suited for longitudinal monitoring of patients with cancer predisposition syndromes or those who belong to high-risk groups requiring vigilant follow-up. However, long acquisition times, quality control issues, cost, and accessibility limit the potential role of MRI as a surveillance tool.

Technical innovations have effectively addressed previous limitations: Fast MRI protocols are now feasible, as nicely presented in the article by Dong and colleagues in this issue of MAGNETOM Flash [54], and abbreviated sequences have been developed specifically to maximize detection sensitivity within constrained timeframes, making MRI surveillance more cost-effective and accessible than ever before. Abbreviated MRI protocols for the breast, liver, pancreas, and prostate exemplify the pragmatic approach now possible: maintaining diagnostic accuracy while dramatically improving workflow efficiency [55–58]. The ability to obtain comprehensive imaging in minutes rather than an hour has fundamentally changed the clinical utility of MRI. Further, it can be used to monitor low-grade malignancies such as small renal masses and low-grade prostate cancers. Some patients with DCIS are under active surveillance, and MRI is a viable option for monitoring this condition [59–62]. This expanding role in both monitoring of indolent tumors and surveillance of high-risk groups will create a significant surge in demand for MRI across healthcare systems worldwide.

The sustainability imperative

As our technical capabilities expand, we must also confront the environmental impact of advanced imaging. The radiology department consumes over 4.2% of a hospital's total energy, and MRI scanners are the most energy-intensive medical devices in radiology [63]. In this issue of MAGNETOM Flash, the contribution by Küstner and colleagues on energy efficiency⁴ in MRI scanners addresses a critical concern about the sustainability of MRI [64]. Their work reminds us that innovation must be measured not only by diagnostic capability but also by resource utilization.

The growing interest in fast scanning and protocol optimization represents a practical response to these sustainability concerns, as they consume significantly less energy while maintaining diagnostic utility for many applications. DL application led to a 72% reduction in energy consumption, and activating Eco Power Mode during non-scanning time resulted in a 30% reduction in energy consumption [63]. Furthermore, low-field MRI such as 0.55T MRI with the MAGNETOM Free.Max system offers another approach to achieving sustainability, as high field strengths consume more energy [65]. Concerns about image quality have been mitigated by applying DL reconstruction, producing comparable results in shorter acquisition times [15]. This evolution toward “right-sizing” our technical resources to clinical needs reflects a maturation of the field, acknowledging that the highest field strength is not always necessary for every clinical question.

Looking ahead: From state-of-the-art to standard of care

As we assess the current state of body MRI, what emerges is not a story of technical compromise but rather one of technological maturity: The tools that once required specialized expertise and were reserved for academic centers have become accessible to community practices. Protocols that once demanded an hour of scanner time can now be completed in minutes. Images that once required expert interpretation can now be enhanced and analyzed with AI assistance, as Hallinan et al. describe in this issue [66]. Some may perceive the transition from 3T to 0.55T, from full protocol to abbreviated protocol, as a downgrading of MRI technology. However, this perspective fails to recognize the strategic advantages these changes offer.

I believe that this democratization of advanced imaging represents the true measure of our progress. Body MRI has not been downgraded from a state-of-the-art modality to a clinical workhorse. It has been elevated from an occasional specialist tool to an everyday clinical necessity. This transition reflects the highest level of technical development: technology so refined it becomes essentially transparent to the user.

As we look to the future, the demand for MRI continues to increase across multiple fronts. We need cutting-edge MRI systems such as the 7T MAGNETOM Terra.X to explore the limits of diagnostic challenges in our new era of personalized medicine. We also need MRI as a problem-solving tool at the highest specifications that are clinically achievable, such as the 3T MAGNETOM Cima.X. Additionally, we require MRI for “the new kids on the block” – screening and surveillance applications, or point-of-care uses – which may not need superb high-end scanners but rather widely available and maintainable technology. The emergence of low-field MRI systems such as the 0.55T MAGNETOM Free. Max is another promising frontier, offering viable options for surveillance and screening applications while significantly expanding access in low-income countries, further democratizing this essential technology.

DL has become essential throughout the entire MRI workflow, from generating high-quality images to automated interpretation. Automatic image quality assessment improves recall process in clinical practice, and automatic segmentation aids quantitative information without additional workload [67–71]. In addition, as whole-body imaging and screening MRI applications increase, radiologists face growing workloads that outpace workforce growth. DL algorithms for risk assessment, already validated for prostate and breast cancer detection, offer a promising solution to this challenge [68, 72, 73]. These tools not only extend radiologists’ productivity but also enhance diagnostic completeness by assisting with non-specialized interpretations. For instance, a body radiologist reviewing images of oncologic patients with back pain for metastases might benefit from specialized algorithms² such as those developed by Hallinan and colleagues for detecting incidental spinal stenosis [66], ensuring patients receive comprehensive care beyond the primary diagnostic focus.

Alongside this expansion, sustainability has become an imperative consideration, with MRI systems still the most energy-intensive devices in radiology departments. DL applications are proving instrumental in addressing this challenge, enabling shorter scan times with comparable image quality, and facilitating synthetic image generation that reduces the need for additional sequences. Low-field systems, once considered a compromise, are now being recognized as an environmentally responsible choice when paired with advanced reconstruction techniques. It provides clinically acceptable images while consuming significantly less energy⁴.

In this evolving landscape, the focus will continue to shift from technical novelty to implementation for clinical utility, from what is possible to what is practical, and from isolated imaging findings to integrated diagnostic

solutions, and finally theranostics. The articles in this issue of MAGNETOM Flash reflect not just the technical advances but also a roadmap for continued evolution, where technical excellence supports real-world clinical practice beyond research.

In this context, body MRI is reaching its full potential as a sophisticated diagnostic tool transformed by technical refinement into an indispensable clinical resource. This transformation is not a compromise, but a triumph – the ultimate goal of all medical innovation.



Jeong Hee Yoon

Abbreviations

CAIPIRINHA	Controlled aliasing in parallel imaging results in higher acceleration
DCIS	Ductal carcinoma in situ
DL	Deep learning
DWI	Diffusion-weighted imaging
FAPI	Fibroblast activating protein inhibitor
GRASP	Golden-angle radial sparse parallel
MET-RADS-P	MEtastasis Reporting and Data System for Prostate Cancer
MY-RADS	Myeloma Response Assessment and Diagnosis System
ONCO-RADS	Oncologically Relevant Findings Reporting and Data System
RF	Radio-frequency
PET	Positron emission tomography
PSA	Prostate-specific antigen
PSMA	Prostate-specific membrane antigen
SAMER	Scout accelerated motion estimation and reduction
SAR	Specific absorption rate
SNR	Signal-to-noise ratio
TE	Echo time
VIBE	Volumetric interpolated breath-hold examination
WB-MRI	Whole-body MRI
XD-VIBE	Extra-dimensional volumetric interpolated breath-hold examination

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¹ MR scanning has not been established as safe for imaging fetuses and infants less than two years of age. The responsible physician must evaluate the benefits of the MR examination compared to those of other imaging procedures.

² Work in progress. The application / the system is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

³ Abdominal imaging at ultra-high-field MRI is a research application and is not released for clinical use. It is not for sale in the U.S. and in other countries and its future availability cannot be ensured.

⁴ The statements for energy and cost savings described herein are based on results that were achieved in settings possible due to certain standard and optional features available on MAGNETOM systems. Because there is no "typical" MRI suite or installation and many variables exist there can be no guarantee that customers will achieve the same results.

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Open Sequence: An Introduction

Thomas Kluge and Christoph Forman

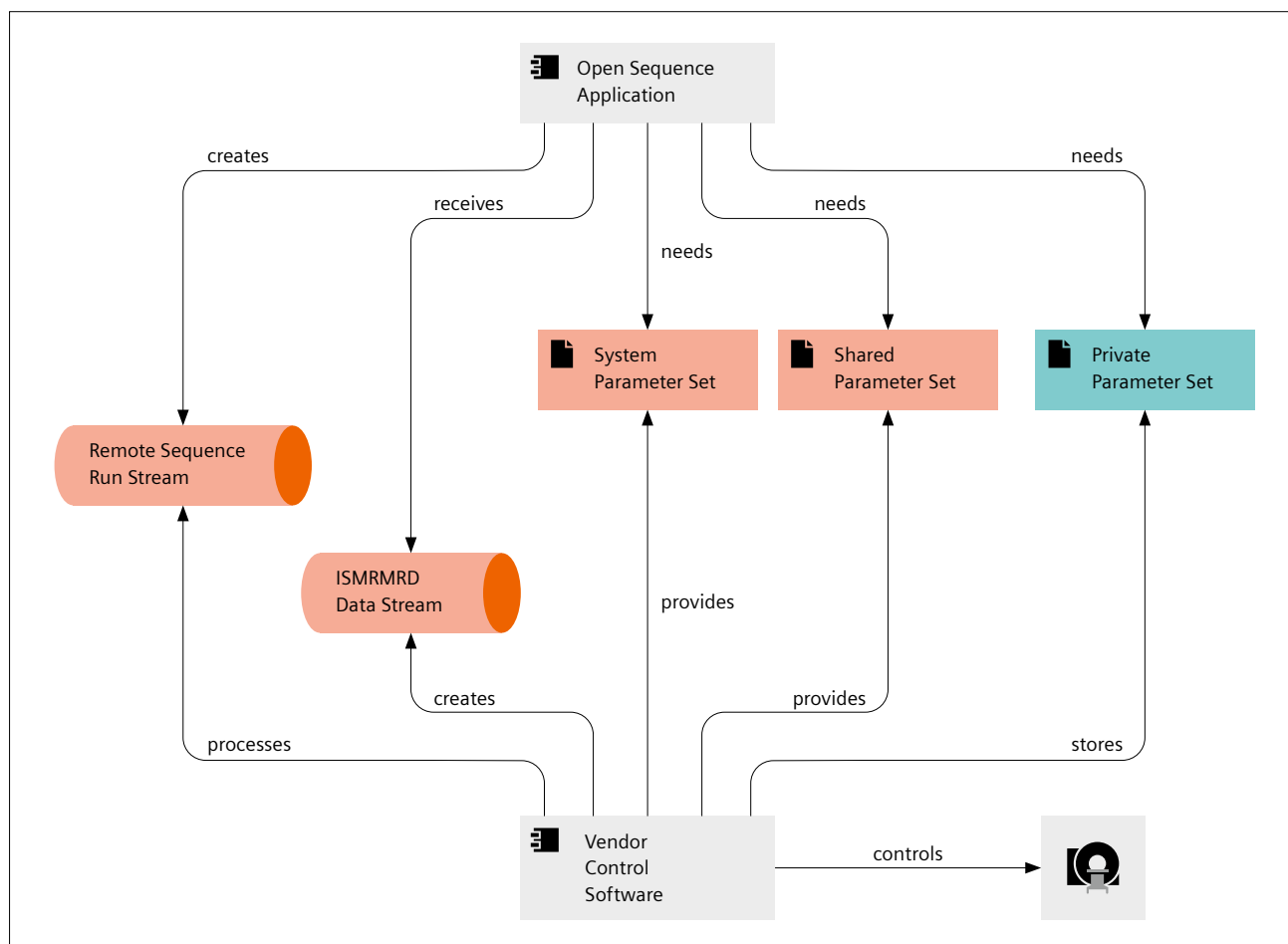
Siemens Healthineers, Erlangen, Germany

Background

In recent years, open innovation and reproducible research have become increasingly important in the research community. For image reconstruction, significant efforts have been made to establish a vendor-independent standard for the storage and transfer of MRI raw data (the ISMRMRD format) [1]. Reconstruction algorithms supporting this format can be distributed, enabling reproduction of results

and application to new data. This can be performed offline for research purposes. On MAGNETOM scanners from Siemens Healthineers, these algorithms can be integrated into the clinical workflow at the scanner using containers and the Open Recon interface.

Several MR sequence development frameworks have been proposed for data acquisition [2–6]. While these



1 The Open Sequence application¹ overview connects an Open Sequence application with the vendor control software. Via the interface, the MR application is parameterized before it provides the MR sequence as a Remote Sequence Run Stream. The acquired data is sent back as an ISMRMRD data stream.

¹Open Recon is to add clinical reconstructions to the system, if signed and released for clinical use by Siemens Healthineers. Any other recon used (e.g., by researchers) is automatically labeled with "not for diagnostic use", which may require observation of national regulations.

Open Sequence is work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

environments claim vendor-independence, the integration on the scanner requires a vendor-specific adapter module provided with these environments. This module is deeply integrated into the vendor's scanner control software and must be adapted and re-released for each software update. This creates significant work for the developers and users of such an environment. To address these shortcomings and to fully support reproducible research, a software-version- and programming-language-independent interface for data acquisition on MAGNETOM scanners has been introduced [7].

The interface was announced at this year's ISMRM Annual Meeting. It will be released with the Open Sequence product on MAGNETOM scanners, providing all necessary interfaces to develop an application for MR data acquisition.

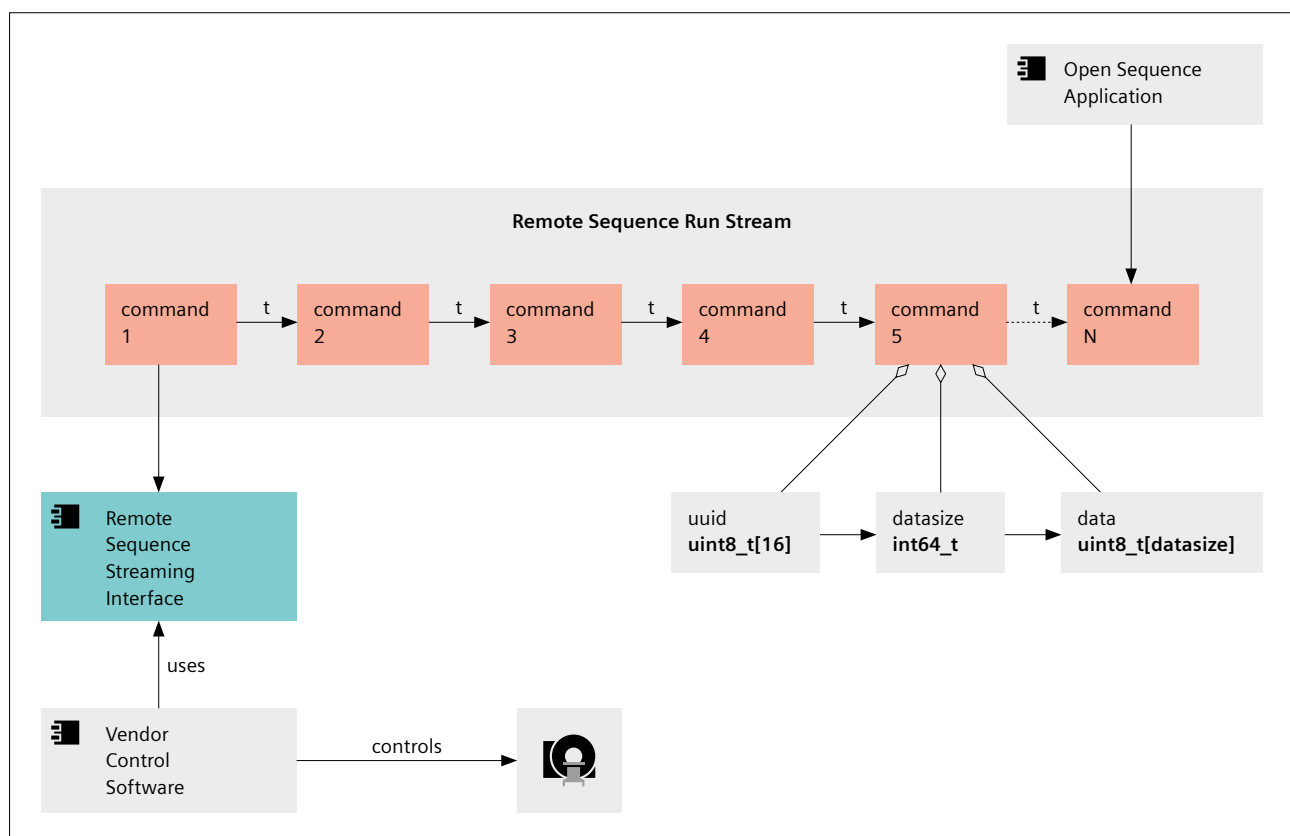
Open Sequence is primarily designed for those who maintain MR sequence programming environments, for researchers, and for developers of MR sequences. Eventually, it will provide physicians, researchers, and patients with access to a wide variety of MR sequence applications that will improve open innovation, reproducible research, and clinical diagnosis. While the interface physically decouples the MR application from the MR scanner and its

control software via network communication, it enables the development of the MR sequence application in any programming language and with any sequence development environment or library. Furthermore, the Open Sequence interface allows to deploy the sequence application without the need to install any binaries on the MR scanner. The Open Sequence interface offers long-term stability and enables the deployment and execution of sequences on every supported MAGNETOM scanner without the need to modify the sequence application.

This article will provide an overview of the main components of the Open Sequence interface, and will then describe use cases for MR applications.

The Open Sequence interface

As illustrated in Figure 1, the Open Sequence interface has a central role by connecting the vendor control software to an Open Sequence application. The interface consists of three main components that are required to describe the MR application: Parameterization, Remote Sequence Run Stream, and the ISMRMRD Data Stream. These components will be discussed in more detail in the following.



- 2** The MR sequence running on the vendor's scanner is defined by the Open Sequence application via a stream of commands. As shown for Command 5, each command has the same basic structure, i.e., a UUID identifying the command, the size of the data associated with the command, and the command content data.

Parameterization

The parameters required by the Remote Sequence are provided by three data structures: the System, Shared, and Private Parameter sets. The System Parameter Set contains parameters describing the MR system, e.g., gradient maximum amplitude and minimum rise time. The Shared Parameter Set provides parameters defined by the user interface of the MR scanner during the examination, e.g., position and orientation of the field of view. Finally, the parameters specific to the MR application are provided in the Private Parameter Set. These private parameters are stored by the MR scanner software and sent to the Remote Sequence without further interpretation by any software component of the MR scanner.

Streaming the MR sequence

As shown in Figure 2, the scanner control data is streamed as a series of commands that are each identified by a unique identifier (UUID) and remain stable across all software versions. With this design, new commands can be added later without the need to modify the interface and without rendering existing Remote Sequences invalid. Each command can be serialized and deserialized in a variety of supported programming languages. Once published, the structure of a single command should never change.

Figure 3 shows an example of such a command embedded in the sequence run stream in detail. The highlighted command represents the start of a time slice within

the sequence diagram and provides its duration. The subsequent commands describe the content of the time slice, finalized by a command ending the time slice. A series of these blocks of commands organized in time slices is used for describing the complete MR sequence. Following this concept, the Remote Sequence Run Stream contains a complete serialized version of the MR sequence that is streamed to the MR scanner for execution.

ISMRMRD data stream

The data acquired by the MR scanner is streamed back to the Remote Sequence using the ISMRMRD format [1]. The Remote Sequence can implement image reconstruction or further process the acquired data.

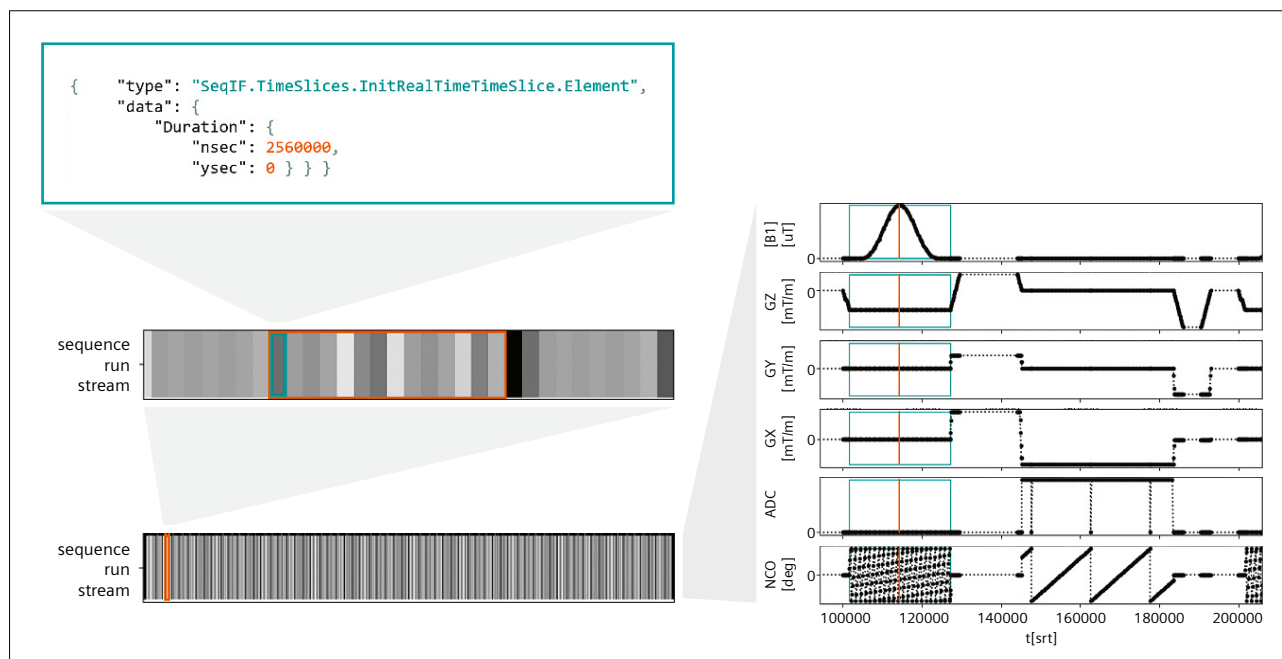
Use cases for MR applications

The Open Sequence interface enables a variety of options for developing and deploying an MR application. In the following, two use cases are introduced.

Researcher plays MR sequence from file

This use case is interesting for the researcher or a student who want to run their – potentially very first – MR sequence on a real MR scanner.

The creation of the Remote Sequence Run Stream and the reconstruction of the data acquired is performed by software written by the researcher in the programming



3 Example of a Remote Sequence command stream for a FLASH sequence. The highlighted command represents the start of a time slice within the sequence diagram and provides its duration. The subsequent commands describe the content of the time slice, followed by a command ending the time slice. A series of these commands is used to describe the complete sequence diagram.

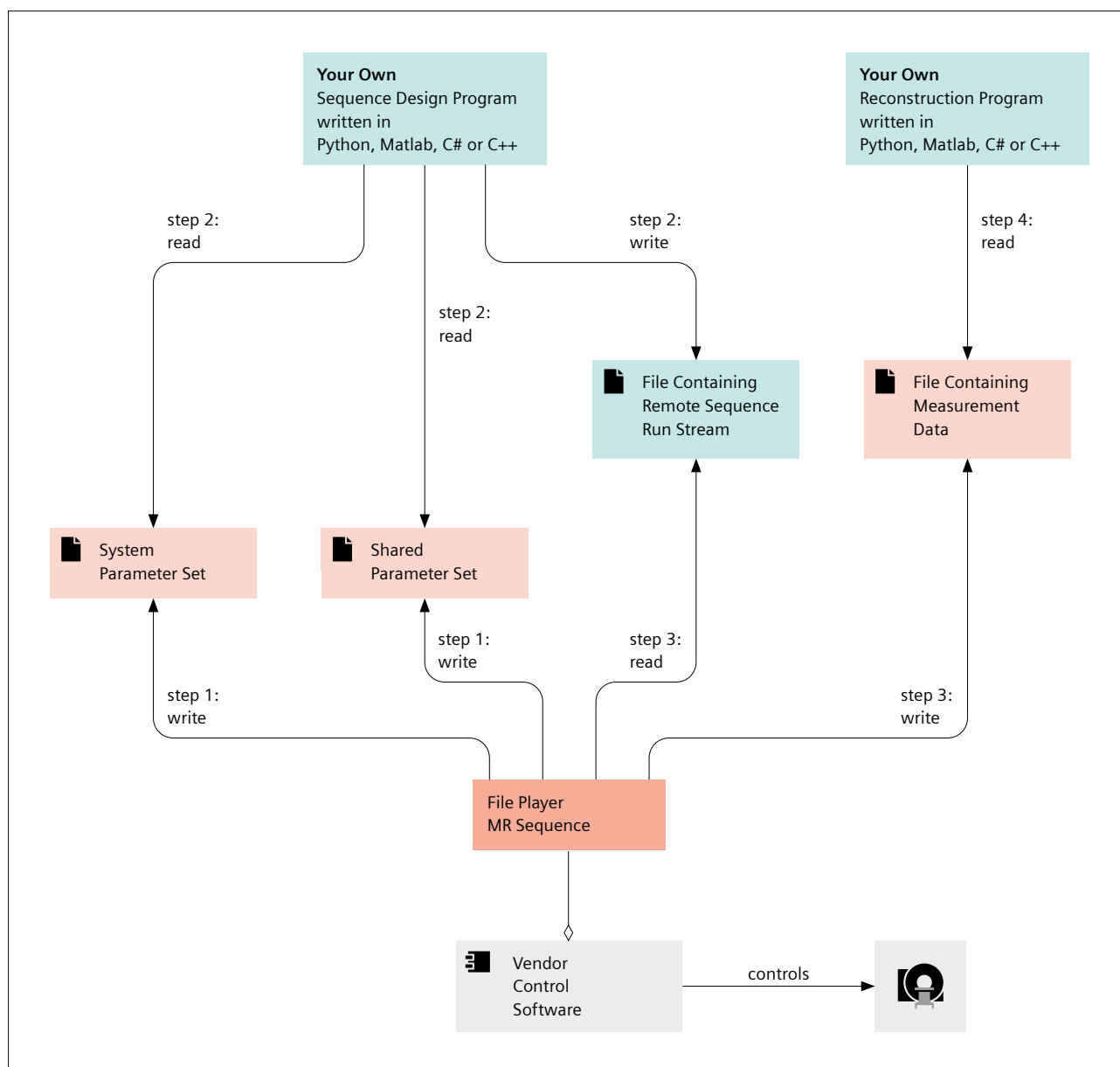
language of their choice. On the MR scanner, a special File Player MR sequence can be used for accessing system and shared parameters, as well as for executing the MR sequence and retrieving the acquired data as a file. The individual steps of this approach are illustrated in Figure 4.

This approach easily provides sufficient scanner access for acquiring data for scientific publications, and supports reproducible research as the files can easily be shared between researchers.

The first activities for connecting the widely adopted Pulseseq [5, 6] framework, in particular the Python version [12], to the MR scanner without using a binary adapter module are based on this use case and demonstrated in [11].

Use case: MR sequence streamed via network interface

In the long term, this use case is aimed at the maintainers of sequence development environments and developers of clinical MR applications.



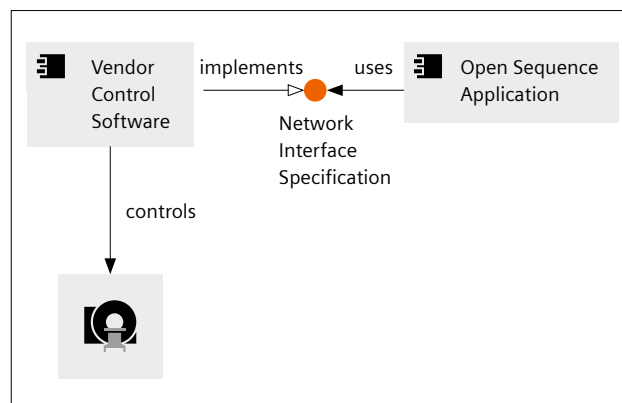
4 File Player use case.

Step 1: Export of the parametrization. Step 2: Creation of the MR sequence as a Remote Sequence Run Stream file. Step 3: Execution of the MR sequence and export of the acquired data. Step 4: Image reconstruction.

In this use case, the MR application is completely decoupled from the MR system via the Open Sequence network interface. It aims to develop and deploy MR applications independent of the software version of the MR scanner, while the MR application can be seamlessly integrated into the examination workflow on the MR scanner.

Based on the documentation of the Open Sequence network interface, the MR application can be developed in any programming language. In the first version of Open Sequence, the MR application is hosted on a computer connected to the MR system via a network cable.

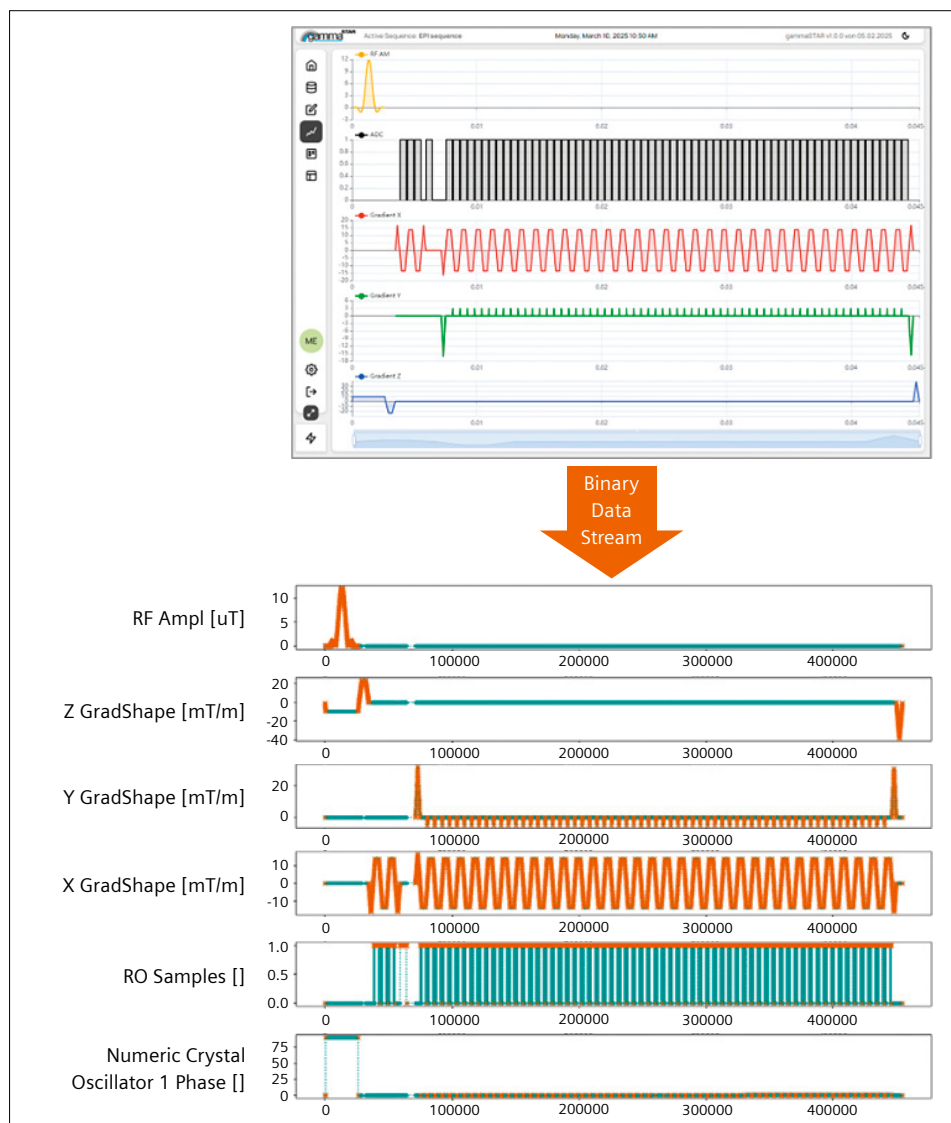
This use case can already be demonstrated by the integration of the gammaSTAR framework [3, 4], developed by the Fraunhofer Institute for Digital Medicine MEVIS. Figure 6 shows an echo planar imaging (EPI) sequence streamed from the gammaSTAR framework to Open Sequence. This sequence has successfully been run on an MR scanner with gammaSTAR running on an external computer.



5 Network streaming use case.

The MR application is physically decoupled via the Open Sequence Network Interface. All communication between the vendor control software and the Open Sequence application is defined by this interface.

6 Example of an EPI sequence defined in gammaSTAR (top), which is streamed via the network interface to Open Sequence and plotted (bottom).



Next steps

The first version of Open Sequence is limited to the execution of predefined streams that have been completely transferred to the control software of the MR scanner before starting the execution. Consequently, the duration of the MR sequence could be limited by the amount of real-time compliant memory available to the control software, and the MR sequences cannot be dynamically defined by reacting to, e.g., physiological events or user interaction. This problem is in the process of being addressed and is described in [10].

The Remote Sequence Streaming Interface has a strong focus on software architecture. However, it is heavily related to work dealing with the question of how to efficiently and effectively develop MR sequences in general. The concept of time slices as small building blocks for MR sequences in particular plays an important role in the MR Fingerprinting Development Kit introduced in [8] and [9].

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Conclusion

The Open Sequence interface enables sequence development independent of the software version of the MR scanner. A wide range of development and deployment opportunities are enabled. Using established sequence development environments like gammaSTAR and Pulseq with the Open Sequence interface further supports open innovation and reproducible research.

Future versions will focus on enabling real-time MR applications and simplifying MR sequence development.

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Optimizing Energy Efficiency in MRI Scanners

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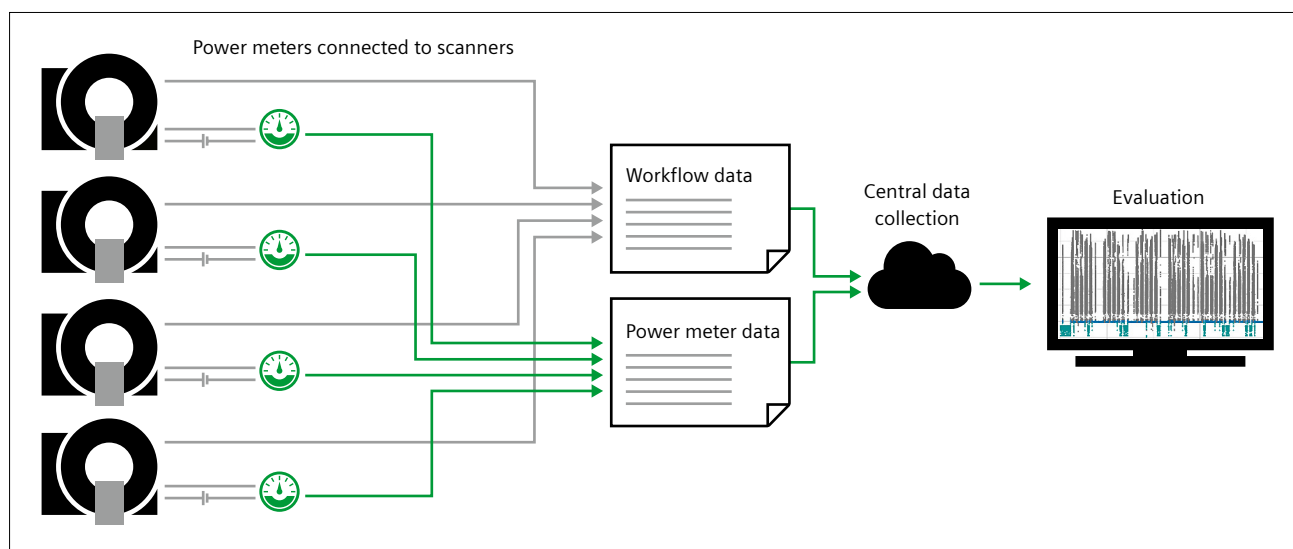
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Introduction

MRI scanners are essential in clinical diagnostics, but they are also among the most energy-intensive devices in healthcare [1, 2]. Their energy consumption has become a significant concern, particularly in light of rising energy costs in Europe, where the price more than doubled between December 2020 and December 2021 [3]. Radiology departments, which account for at least 4.2% of a hospital's total energy consumption, are major contributors to both energy costs and greenhouse gas emissions [1, 4–8]. Within these departments, MRI scanners consume more than 100 megawatt-hours (MWh) per year on average, making them the highest energy consumers compared to other equipment [4, 9]. As demand for MRI devices grows, finding ways to optimize their energy efficiency and operate them sustainably has become a pressing priority.

The primary drivers of MRI energy consumption are the magnet cooling system and the gradient system, with magnet cooling alone accounting for approximately 42% of total energy use [10]. Active scanner operation demands more energy than idling [6], with productive use accounting for up to 70% of the total energy consumed [9]. Hence, reducing energy consumption during active operation or shortening the duration of active scanner operations offers the most promising benefits. One way to achieve this is by shortening acquisition times, though maintaining diagnostic quality remains a critical consideration here. Optimizing protocols and incorporating artificial intelligence (AI)-accelerated sequences (undersampled acquisitions paired with AI-based image reconstructions) can contribute to energy savings. In this regard, AI-based reconstructions



1 The MRI scanners are connected to power meters with a 1-hertz (Hz) sampling rate. Synchronized power-meter, sequence, parameter, and examination data are collected and centrally stored. Eventually, insights into workflow are gained and the impact of individual scan parameters on the scanner's power consumption is evaluated.

of undersampled or noisy data have demonstrated the ability to reconstruct high-fidelity images without compromising image quality [11–13]. These innovations thus provide promising opportunities to balance energy efficiency with reduced scan times. On the other hand, reducing acquisition times may result in the freed-up time being used for increased patient throughput, which would diminish the energy savings. Hence, it is critical to observe the complete workflow and propose energy optimizations along the whole imaging chain.

In a first study [10], we therefore investigated and analyzed energy consumption on two MRI scanners in a musculoskeletal setting when using optimized protocols and AI-accelerated sequences. Our findings showed that it was possible to achieve significant energy savings by optimizing protocol settings and implementing AI technologies. The outcomes motivated us to collect a more long-term and comprehensive database. We have therefore installed automatic energy measurement devices (power meters) in a fleet of MRI scanners (currently five scanners, with more being prepared) at our hospital to monitor the energy consumption behavior in a clinical setting during regular MRI workflows. We aim to identify the specific energy demands across different scanner models, operational modes, and imaging sequences. The findings will provide insights into which features drive energy consumption and where substantial efficiency improvements can be made. The focus will be on energy-saving optimizations, such as protocol refinement, AI-accelerated imaging, and cooling-system adjustments. Ultimately, the goal is to optimize the complete MRI workflow to enable more sustainable MRI operations while maintaining the high standards required for diagnostic quality.

Energy measurement

We are collecting and analyzing extensive data on energy consumption and scanner use from our primarily clinical MRI scanners, creating one of the most detailed datasets available to date. For this first-of-a-kind installation (Fig. 1), data collection began in June 2024 and is ongoing. Each scanner is equipped with a Siemens SENTRON 7KM power meter (Siemens AG, Chem, Germany) that allows continuous and precise energy tracking. General information, including the time of the measurements, scan duration, and the operated scanner, were collected. Further, detailed information such as the system and scan parameters for each sequence were documented. In total, 122 parameters are recorded, capturing detailed information that enables a granular analysis of power use relative to imaging requirements. Although not all parameters are relevant to every sequence, this exhaustive dataset allows to identify

features driving energy consumption and facilitates in-depth correlation analysis between power consumption and specific protocol settings. This data on the operational use of MRI scanners is logged and automatically collected. Each piece of operational use data is time-synchronized and associated with its specific power consumption data.

We are interested in obtaining the following:

1. Observational insights into workflows and energy consumption, especially the influence of certain parameters
2. Insights into the potential benefits of AI-accelerated imaging to save energy
3. An estimate of potential power savings by identifying idle periods in which a scanner is “ready for measurement” but remains inactive (with high power demand)

We consider these extended idle periods to be potentially suitable for activating Eco Power Mode (switching off the helium pump during off-cycles and reactivating it based on temperature and pressure values), which could reduce energy consumption by optimizing the magnet cooling. Idle intervals longer than one hour, where Eco Power Mode was not activated, are flagged as opportunities for additional savings.

Findings

A first study focused on two scanners in a practice setting and investigated musculoskeletal imaging. The results indicate that reducing scan time significantly contributes to energy savings [10]. We found that, across all body regions, the average reduction in energy consumption was 72% with AI-accelerated sequences, and 31% with economic protocols (shortened imaging sequences using slightly increased acceleration factors or reduced resolution). This was accompanied by time reductions of 71% (AI) and 18% (economic protocols) compared to a conventional baseline. Optimizing the cooling system using Eco Power Mode during the non-scanning time showed a 30% lower energy consumption. In detail, we observed an average scan time reduction for optimized (economic) protocols of 34% (hip: 235 s, $p < 0.001$), 32% (knee: 148 s, $p < 0.001$), 11% (spine: 57 s, $p < 0.01$), and 25% (shoulder: 146 s, $p < 0.01$) compared to the baseline. AI protocols in turn resulted in an average scan time reduction of 81% (hip: 467 s, $p < 0.001$), 66% (knee: 268 s, $p < 0.001$), 49% (spine: 218 s, $p < 0.001$), and 77% (shoulder: 367 s, $p < 0.001$) over the baseline. This resulted in an average energy reduction of 21% (hip: 1.9 kilowatt-hours (kWh), $p = 0.31$), 48% (knee: 4.0 kWh, $p < 0.001$), 22% (spine: 2.3 kWh, $p < 0.01$), and 28% (shoulder: 2.9 kWh, $p < 0.01$)

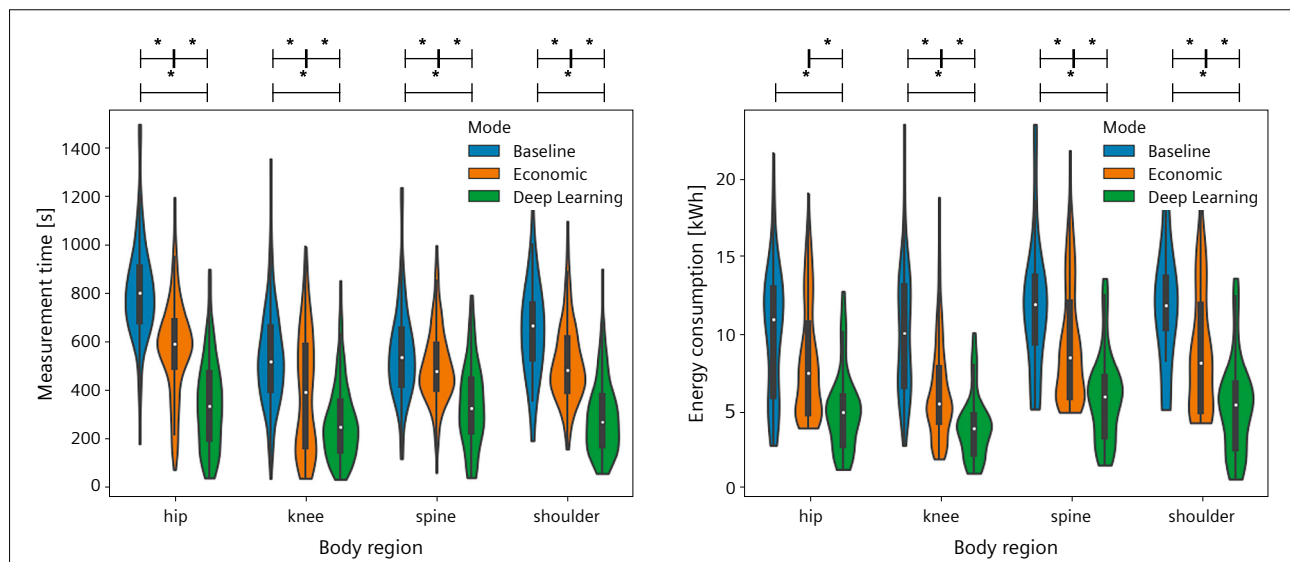
for optimized protocols over baseline, and a reduction of 69% (hip: 5.2 kWh, $p < 0.001$), 89% (knee: 6.3 kWh, $p < 0.001$), 66% (spine: 5.8 kWh, $p < 0.001$), and 76% (shoulder: 6.5 kWh, $p < 0.001$) for AI protocols over baseline. The image quality of the optimized and AI-accelerated sequences was comparable to the baseline. Our results are depicted in Figure 2. Considering 250 working days with 10 hours of standby overnight, and 115 days without operation, this reflects an average energy saving per scanner and year of 14.1 MWh, which equates to 5,414 kg of CO₂ (385 g CO₂/kWh in Germany), if AI sequences are used. Whilst this is promising, several additional variables, such as sequence, application, and targeted body region, will impact the total energy consumption. To better understand these individual influences and to propose an evidence-driven parameter adjustment for energy reduction, we set up a second, larger-scale study.

In this ongoing study, we are currently analyzing data gathered during clinical examinations on five of our clinical MRI scanners, on which a variety of protocols are performed. An example of energy consumption over a three-month period is depicted in Figure 3. The most energy is consumed during the scanner's productive (non-idle) times (Fig. 3, top). Across the five scanners, we identified that activating Eco Power Mode during idle periods could yield savings of up to 48.7 MWh yearly, equating to 18,750 kg of CO₂. As expected, scan duration significantly influences overall energy consumption (Fig. 4). However, the scan duration explains only 57% of the total variance in energy usage,

indicating that other factors also influence energy consumption. Additionally, the field strength (1.5T vs 3T) has a considerable impact, with the 3T scanner consuming noticeably more energy (Fig. 4). The selected protocols (and the sequences they contain) have a substantial impact on energy consumption, with savings of up to 1 kWh per applied protocol. For nearly 1,919 unique protocols, we generalized and simplified them to identify some common behaviors. Adjustment sequences require little energy, but their frequent occurrence within the protocol can easily accumulate to a significant amount of energy use. Furthermore, the slice thickness influences energy consumption, with thinner slices resulting in higher energy use. Spin-echo sequences show a higher energy demand than gradient-echo sequences (Fig. 5). However, it should be noted that this is also linked to the imaged body region, for which sequences are optimized, and which subsequently have a different scan duration footprint (for example, comparing T1w TSE of several minutes in the brain versus single breath-hold Dixon imaging of a few seconds in the abdomen). For a comprehensive analysis, we thus need to include all features to find the best workflow optimizations to reduce energy consumption.

Conclusion

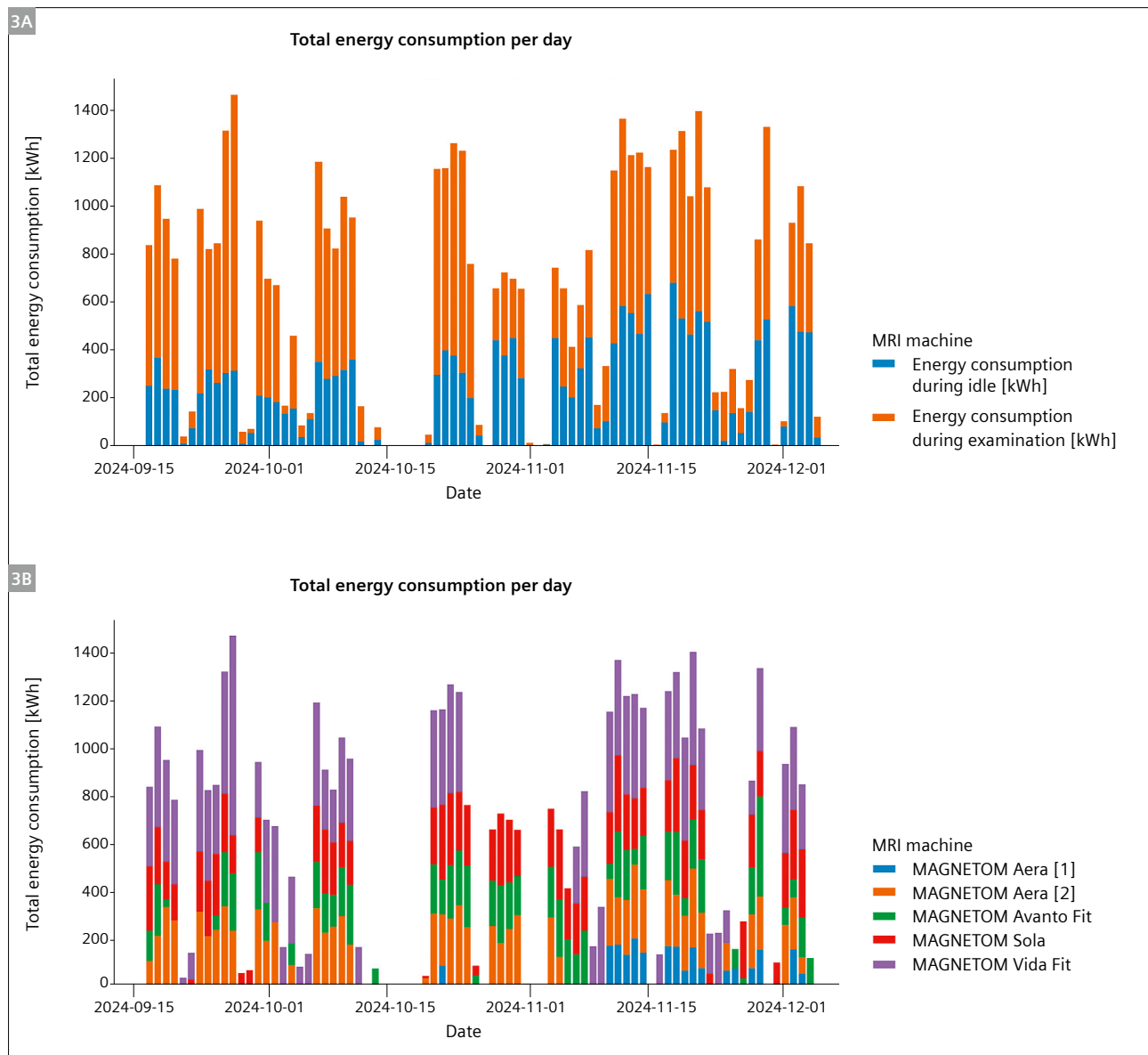
Our studies are the first to show in detail, in a fleet of MRI scanners over several months, the energy consumption correlated to MRI sequences and various features. We high-



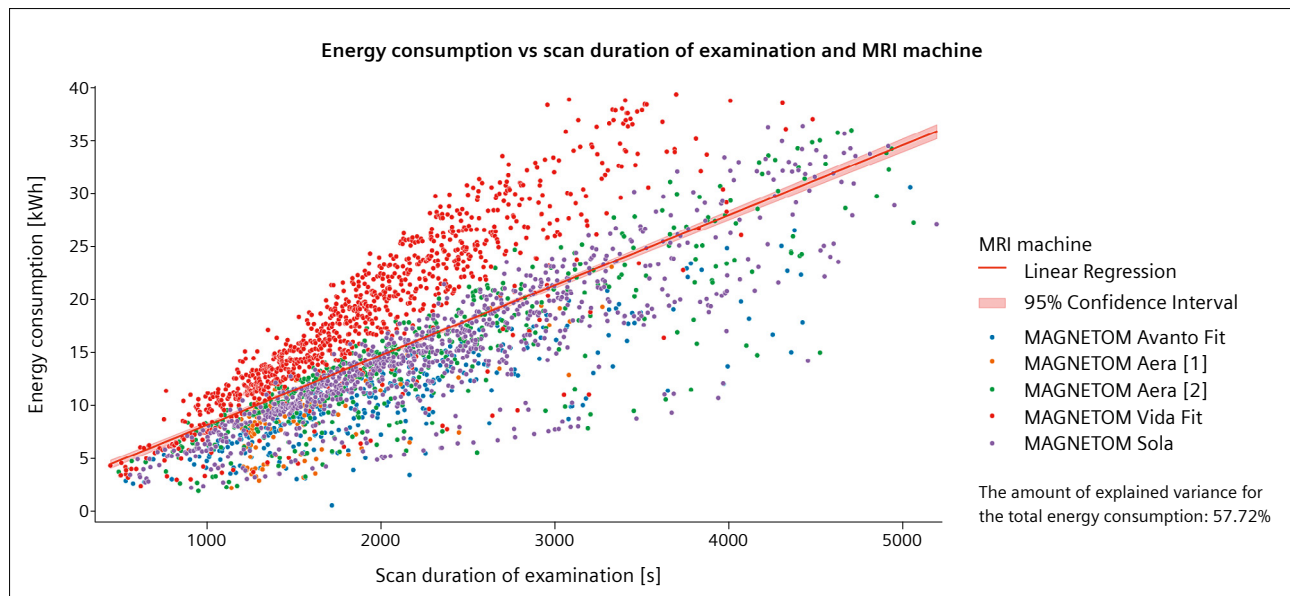
2 First study: Acquisition times (left) and energy consumption (right) for each body region (hip, knee, spine, and shoulder) of the musculo-skeletal protocols in two MRI scanners. Energy consumption modes are depicted: baseline (no modifications), economic (shortened sequences), and deep learning (AI-accelerated sequences). Data distribution, mean (white dot), interquartile range (solid black vertical box), and standard deviation (thin black vertical bars) are shown. Statistical significance with a p-value of 0.05 is indicated by the asterisks.

light the potential of Eco Power Mode and of AI-accelerated sequences to significantly reduce energy consumption. The thorough analysis of features such as scanner type, scan duration, protocol, and scan parameters is the first step toward identifying factors contributing to energy consumption. In the future, these findings will enable us to improve energy efficiency and recommend workflow optimizations

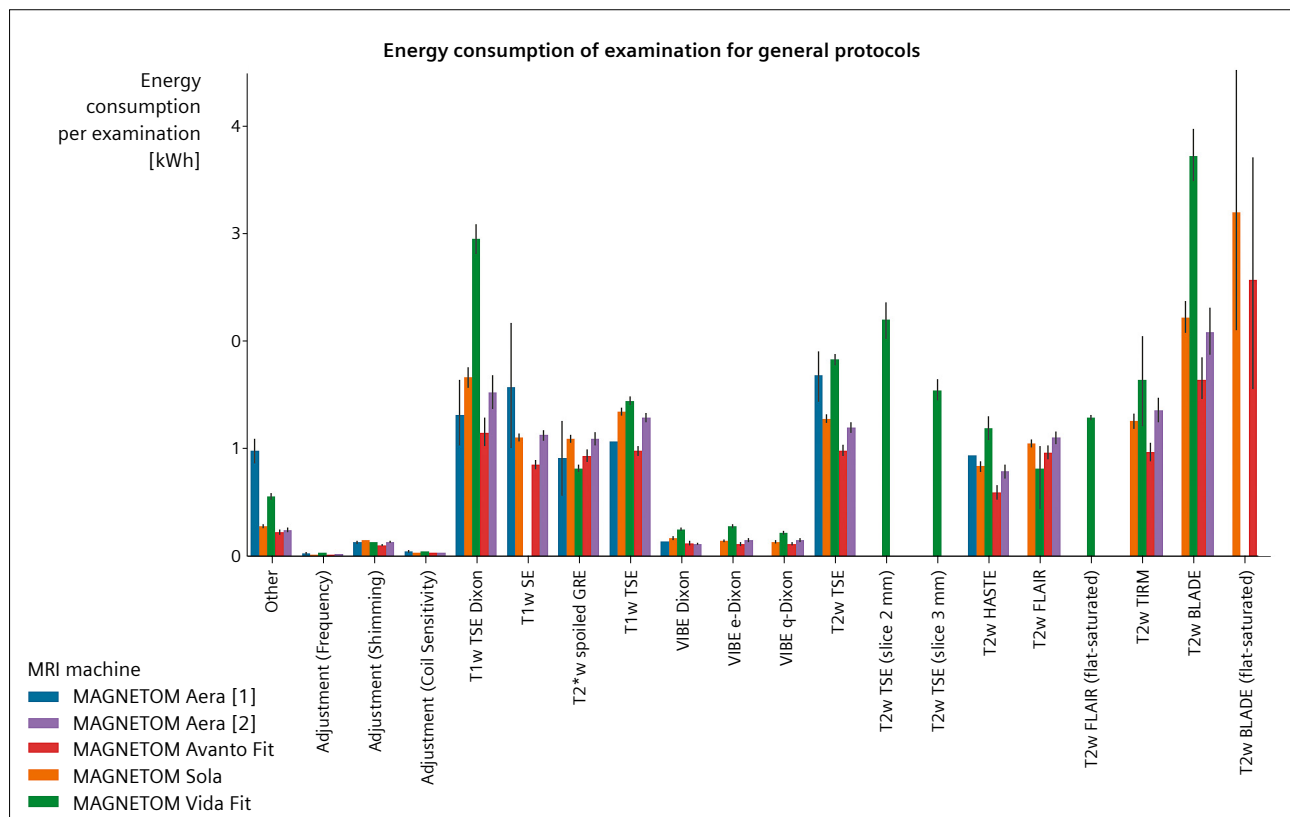
to reduce energy consumption. In addition, a closer examination of the interactions between the features is necessary to gain a deeper understanding and develop effective trade-offs. Overall, implementing these strategies will drive substantial energy and cost savings, and will support more sustainable MRI operation in clinical settings.



3 Second study: Example depiction of total daily energy consumption over three months. **(3A)** Total energy consumption divided into energy consumption during idle and productive (non-idle) states. The total energy consumption is the sum of the idle and productive states. **(3B)** Total energy consumption of the different MRI scanners.



- 4** Second study: The scan duration of the entire examination/protocol shows a positive relationship with overall energy consumption. However, the scan duration explains only 57% of the total variance, indicating that other factors also influence energy consumption. The plot reveals two distinct groups: The MAGNETOM Vida Fit, a 3T scanner (Siemens Healthineers, Erlangen, Germany), exhibits a steeper slope in its relationship with scan duration, while the 1.5T scanners have a gentler slope.



- 5** Second study: The harmonized sequences were plotted in relation to their energy consumption. Adjustment sequences generally consume less energy. Sequences with thinner slices require more energy – e.g., T2w TSE (2 mm) vs. T2w TSE (3 mm), where the number in brackets refers to the slice thickness. Spin-echo sequences demand in general more energy than gradient-echo sequences.

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The statements and recommendations for energy and cost savings described herein are based on results that were achieved in settings possible due to certain standard and optional features available on Siemens MAGNETOM systems. Because there is no "typical" MRI suite or installation and many variables exist (e.g., MRI location, site specific installation requirements, case mix, level of IT and/or automation adoption) there can be no guarantee that customers will achieve the same results.

From Concept to Deployment: A Deep Learning Model for Lumbar Spinal Stenosis at National University Hospital in Singapore

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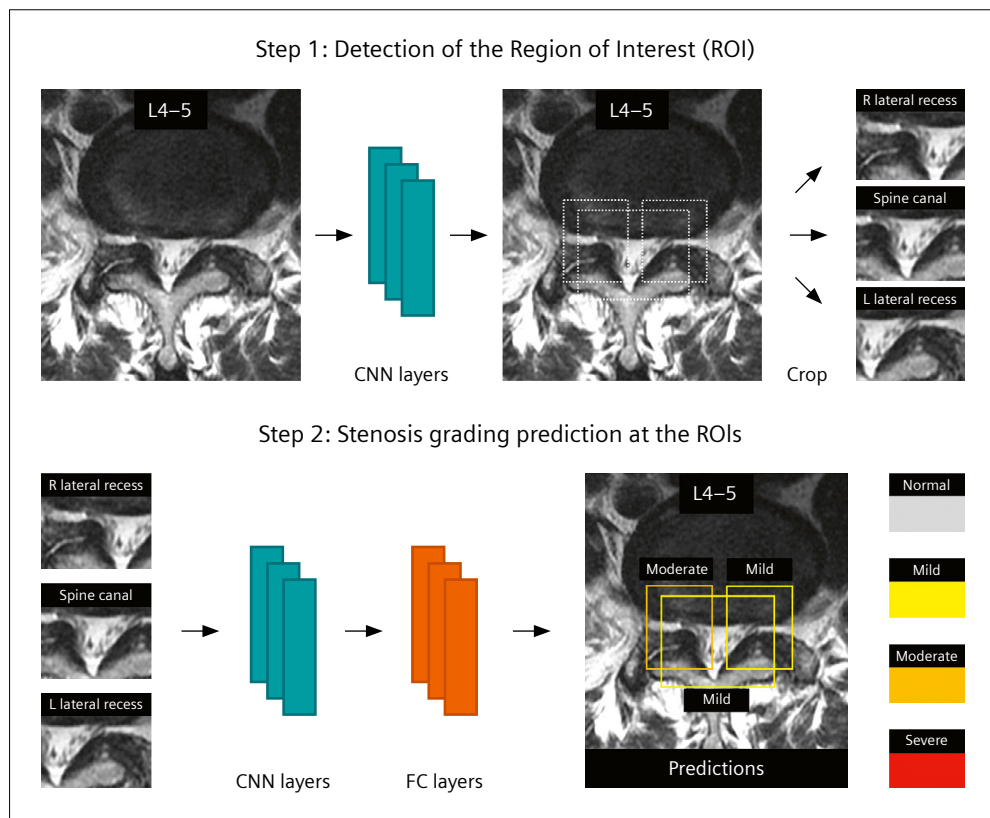
³Siemens Healthineers

Introduction

Lumbar spinal stenosis (LSS) is one of the most common degenerative conditions in adults, with around 200,000 individuals diagnosed annually in the United States [1–3]. Given the pivotal role of magnetic resonance imaging (MRI) in diagnosing and grading LSS, radiologists must routinely evaluate multiple lumbar levels to check for narrowing at the spinal canal, lateral recesses, and neural foramina. Despite the critical role of MRI, reporting LSS remains

a time-intensive, repetitive task for radiologists, often compounded by poor interobserver agreement in stenosis grading. Recognizing these challenges, a multidisciplinary team from the National University Hospital (NUH) in Singapore, the National University of Singapore (NUS), and external collaborators set out to develop a deep learning (DL) solution (Spine AI)¹ aimed at streamlining LSS evaluation.

¹MR Lumbar Spine AI is currently under development and not commercially available. It is for research use only. Not for clinical use. It is not for sale in the USA and its future availability cannot be ensured.



1 Architecture of the CNN-based models for lumbar stenosis. Step 1 involves a CNN that detects the ROI. Step 2 applies another CNN to classify stenosis severity, as illustrated here for the spinal canal and bilateral lateral recesses at the L4–L5 level. Abbreviations: CNN: convolutional neural network, ROI: region of interest, FC: fully connected, R: right side, L: left side

Motivation behind the project

The project was born from the clinical realities faced by our radiology department at NUH, where a typical Monday morning could see over 50 lumbar spine MRIs queued for reporting. The tedious nature of detailing stenosis at the spinal canal, lateral recesses, and neural foramina inspired us to explore whether DL could semi-automate this process and thereby reduce variability and achieve time savings. Additionally, prior studies had demonstrated promise in applying DL to lumbar spine MRI, but none had comprehensively addressed all three regions of interest (ROIs) or integrated explainable outputs to aid radiologists [4, 5].

The development of the Spine AI model was made possible through a multidisciplinary collaboration that included radiologists, computer scientists, and clinicians. Key contributors included Professor Beng Chin Ooi's computer science team at NUS, whose work in advanced machine learning helped refine the AI architecture. The first step in development was securing institutional ethics approval, which paved the way for collecting high-quality imaging data across a range of vendor platforms to create a generalizable DL solution.

Development of the deep learning model

Our initial technical study on the LSS model was published in *Radiology* in 2021 [6]. Over a three-year period (September 2015 to September 2018), the team extracted 446 lumbar spine MRI studies for DL model training and testing. Exclusions included studies with spinal instrumentation, scoliosis, suboptimal image quality, and post-gadolinium sequences. Axial T2-weighted images were used for assessing the spinal canal and paired lateral recesses, while sagittal T1-weighted images were used for analyzing the paired neural foramina. The dataset was split into an internal training set (80%), a validation set (9%), and a test set (11%). External validation included a dataset of 100 lumbar spine MRI studies from Saudi Arabia, to assess the model's generalizability across geographic regions.

Four radiologists with subspecialty expertise in spine imaging (musculoskeletal radiologists and neuroradiologists with 2–9 years of experience) labelled the training data using bounding box annotations. This supervised labelling ensured high-quality inputs for model training. When drawing each bounding box at the region of interest, the radiologist classified stenosis on axial and sagittal MRI based on established criteria and visual scales for the spinal canal [7–9], lateral recesses [10], and neural foramina [7]. Overall, stenosis was categorized into four clinical grades: normal, mild, moderate, and severe. These corresponded to stenosis of 0, < 1/3, 1/3–2/3, and > 2/3 of the normal diameter, respectively. We decided not to use more granular quantitative measures, as these are not typically used in clinical practice [11].

The developed DL model consisted of two main convolutional neural networks (CNN). The first CNN was tasked with localizing ROIs: the spinal canal, bilateral lateral recesses, and neural foramina. The second CNN performed the actual stenosis classification, assigning each region to one of four severity grades (normal, mild, moderate, or severe). This approach bypassed the pitfalls of relying on unstructured radiology reports by requiring high-quality annotated datasets (Fig. 1).

Performance evaluation

The performance of the DL model was evaluated using both an internal test set at NUH (50 cases) and an external test set of 100 lumbar spine MRI studies provided by a collaborator in Saudi Arabia. The internal test set was labelled by a senior musculoskeletal (MSK) radiologist, Professor Hiroshi Yoshioka, who has over 30 years of experience and served as the reference standard. Two subspecialist radiologists, a neuroradiologist with five years of experience and an MSK radiologist with nine years of experience, also labelled the test dataset to enable comparison with the DL model.

When detecting ROIs, the DL model achieved recall rates exceeding 99% for the spinal canal and over 95% for the lateral recesses, which closely matched the performance of the subspecialist radiologists. For the neural foramina, the DL model's recall was 84.5%, comparable to one of the radiologist's recall at 83.9%. This slightly lower performance in foraminal detection likely reflects the challenge posed by less distinct anatomical borders between the foramina, the spinal canal, and the paravertebral region.

Interobserver agreement for stenosis classification was calculated using Gwet's kappa (AC1), which took into account the higher prevalence of normal classifications in the dataset. Under a dichotomous classification scheme (normal/mild versus moderate/severe), the central canal demonstrated almost perfect agreement, with kappa values of 0.98 for both radiologists and 0.96 for the DL model. For the lateral recesses, the model achieved a kappa value of 0.92, aligning with one radiologist's score and slightly lower than the 0.95 for the other radiologist. Neural foraminal agreement for the model was 0.89, which was slightly reduced compared to the subspecialist radiologists' kappa values of 0.95 and 0.94. Notably, on the external test set, the model's performance for the neural foramina improved to 0.96, accompanied by high agreement for the central canal (0.95) and lateral recesses (0.96). This strong performance on external testing suggested the model's potential for broader application across different populations and imaging protocols.

This initial article showed promising results and was accompanied by an editorial in *Radiology* by Professor

Hayashi (2021), which underscored the importance of further evaluating clinical utility [12]. While recognizing the model's technical success, Professor Hayashi stressed the need for studies assessing how AI assistance might reduce interpretation times, standardize reporting, and reduce variability among radiologists. For our team, these insights served as a roadmap for further development and assessment of the DL model.

Clinical validation and key findings

The potential clinical impact of the Spine AI DL model was evaluated in a second study in *Radiology* in 2022, which focused on the model's ability to improve productivity and consistency among radiologists reporting lumbar spine MRI studies [13]. In this retrospective study, eight radiologists with experience ranging from 2 to 13 years (in-training and attending radiologists) interpreted lumbar spine MRI studies both with and without AI assistance. A cross-over design with a one-month washout period was used to minimize learning or recall bias. As before, the radiologists assessed stenosis at three key regions along the lumbar spine: the spinal canal, lateral recesses, and neural foramina. The evaluation included both reporting time and interobserver agreement, using a reference standard which was again provided by Professor Yoshioka, an external MSK radiologist with over 30 years of experience.

The results demonstrated a significant reduction in reporting time when AI assistance was used. On average, interpretation time decreased from around 2–4.5 minutes (124–274 seconds) without AI to 1 minute or less (47–71 seconds) with AI ($P < 0.001$). These time savings were most marked for general and in-training radiologists, who appeared to benefit the most from the automated stenosis gradings and bounding box visualizations provided by the Spine AI model. Specifically, musculoskeletal radiologists reduced their mean reporting time from 124 ± 25 seconds to 47 ± 24 seconds (a 62% decrease), general radiologists

cut their time from 226 ± 61 seconds to 70 ± 29 seconds (a 69% decrease), and in-training radiologists experienced the greatest drop, declining 74% from 274 ± 88 seconds to 71 ± 28 seconds (Table 1).

In addition to improved efficiency, the AI model enhanced the consistency of stenosis grading for non-specialist radiologists (Table 1). General and in-training radiologists demonstrated marked improvements in interobserver agreement for four-class neural foraminal stenosis when using AI assistance, with Gwet's kappa values increasing from 0.39 (without AI) to 0.70–0.71 (with AI). For other stenosis gradings, AI-assisted radiologists achieved either superior or equivalent interobserver agreement compared to unassisted interpretations. These findings highlighted the DL model's ability to bridge the gap in performance between specialist and non-specialist radiologists.

Transition to clinical deployment: The syngo.via Frontier platform from Siemens Healthineers

Having established both the technical feasibility and clinical benefits, the team turned its attention to deployment of the Spine AI model. In November 2023, a technology development agreement (TDA) was signed by NUH, NUS, and Siemens Healthineers to integrate the Spine AI model into the syngo.via Frontier platform as a research application (MR Lumbar Spine AI¹) which will be made available in the marketplace. This marked the most exciting phase in the project so far, effectively taking the DL model from bench to bedside. As many radiologists are aware, seamlessly incorporating a DL model into routine workflows is no simple task, and involves close engineering collaboration, robust user-interface and user-experience (UI/UX) development, and alignment with clinical safety and security protocols.

Collaboration with Siemens Healthineers was essential for providing key advancements in the overall performance

Radiologist Group	Reading Time (sec)		Time saved (%)	Neural foramina (κ)	Lateral recess (κ)	Spinal canal (κ)
	Without DL	With DL		Without DL → With DL		
MSK	124 ± 25	47 ± 24	62	0.83 → 0.87	0.81 → 0.78	0.86 → 0.85
General	226 ± 61	70 ± 29	69	0.71 → 0.87	0.68 → 0.76	0.74 → 0.84
In-training	274 ± 88	71 ± 28	74	0.71 → 0.88	0.65 → 0.77	0.85 → 0.80

Table 1: Summary of mean interpretation times and two-class interobserver agreement for MRI lumbar spine stenoses with and without DL model support

Note: Reading time is presented as mean $\pm$ standard deviation, and time saved (%) reflects the reduction in mean reporting time. Gwet's kappa (κ) values indicate interobserver agreement for two-class stenosis grading (normal/mild versus moderate/severe) without → with DL assistance.

of the lumbar spine DL model regarding their use in a clinical setting: vertebral disc level detection was developed and is now robust, ensuring the correct levels are identified for accurate autoreport generation of stenosis gradings. Using the AutoAlign Spine technology for the MRI scan, can further improve the accuracy of the vertebral disc level detection. The engineers at Siemens Healthineers also helped refine features, such as presenting bounding boxes only at the most prominent levels of stenosis at each ROI, thus reducing clutter and focusing the radiologist's attention where it matters most.

The lumbar Spine AI application is now available as a research prototype¹ on *syngo.via* Frontier. The automatic processing involves carefully mapping MRI sequences to the model's input requirements. When a lumbar spine MRI study is opened, the system automatically identifies axial T2-weighted and sagittal T1-weighted images, then applies the AI algorithms in the background. Color-coded bounding boxes corresponding to different stenosis severities are subsequently overlaid on each relevant level of the lumbar spine. These results are stored in the form of grayscale

presentation state overlays on the source images. The model also generates an automated summary report, which radiologists can review and edit. This can be output as either a DICOM structured report or a DICOM-encapsulated PDF, both of which are designed to merge smoothly into existing radiology information systems. Automated reports, bounding boxes, and stenosis grading examples are provided in Figures 2–4, which show a full range of stenosis gradings, even on a background of scoliosis.

At NUH, deployment of the Spine AI model on *syngo.via* and prospective evaluation are now taking place. *syngo.via* has been installed on all radiologists' desktops, allowing for the incorporation of Spine AI into routine reporting workflows. The Spine AI clinical innovation and deployment has also been supported by multiple grants from the National Medical Research Council (NMRC), including the New Investigator Grant and Clinical Innovator Awards, and it has been recognized through the Practice Changing Innovations Award (NUH and Ministry of Health, Singapore).

2A

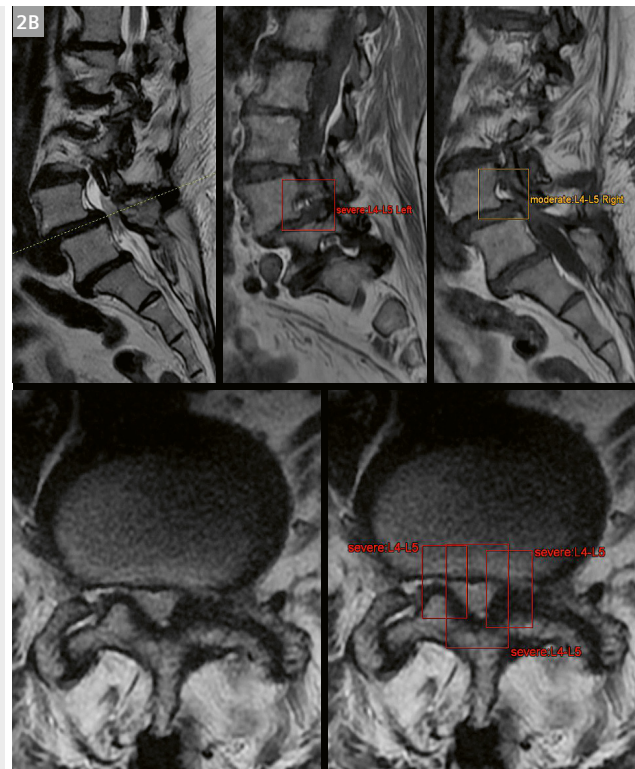
Spine AI report

At T11-T12 there is mild spinal canal, no bilateral recess stenosis. There is no right and mild left neural foraminal stenosis.
At T12-L1 there is mild spinal canal, mild bilateral recess stenosis.
At L1-L2 there is moderate spinal canal, severe right and moderate left lateral recess stenosis.
At L2-L3 there is moderate spinal canal, no right and mild left lateral recess stenosis. There is moderate right and no left neural foraminal stenosis.
At L3-L4 there is no spinal canal, mild right and severe left lateral recess stenosis. There is severe right and no left neural foraminal stenosis.
At L4-L5 there is severe spinal canal, severe bilateral recess stenosis. There is moderate right and severe left neural foraminal stenosis.
At L5-S1 there is mild spinal canal, mild right and moderate left lateral recess stenosis. There is mild right and severe left neural foraminal stenosis.

Report based on automated processing!
Research only, Not for clinical use

Input Data:
5 SAG T1
7 AX T2 TOP
8 AX T2 MID
9 AX T2 BOT

2B



- 2** Spine AI predictions for Patient 1, a 70-year-old woman with back pain and bilateral lower limb radiculopathy. The assessment was limited by scoliosis, although the most severe stenoses in the lower lumbar spine from L3 to S1 were detected and classified correctly. An autogenerated Spine AI report is shown in Figure (2A). In Figure (2B), the top left is a sagittal T2-weighted image showing the scoliosis and spinal deformity with a cross-reference line at L4–5. Sagittal T1-weighted images are provided in the top row and show neural foramina stenosis gradings. Axial T2-weighted images are provided in the bottom row and show the paired lateral recesses and central spinal canal stenosis gradings at L4–5.

Future directions: Expanding capabilities and exploring new frontiers

While the deployment of the Spine AI model is a significant achievement, there is great potential for further development. Future work aims to expand the model beyond stenosis grading to include other pathologies such as spondylolisthesis, fractures, and disc disease. Additionally, the team plans to extrapolate its application to other regions, such as the cervical spine² [14], and explore the deployment of additional tools for detecting spinal metastases² and early epidural spinal cord compression² on MRI and CT imaging, supported through the New Investigator Grant and Clinical Innovator Award from the National Medical Research Council [15–18]. Initial work on these fronts has already been presented at national and international conferences, including the European Society of Skeletal Radiology (ESSR) conference in 2024 and the Global Spine Congress (GSC) in 2023.

²The product/features (mentioned herein) are not commercially available. Their future availability cannot be guaranteed.

Building on the success of the lumbar spine work, the group recently introduced a DL model for degenerative cervical spondylosis in The Spine Journal in 2024 [14]. This transformer-based model was trained on MRI data from 454 patients at NUH, and tested on an internal dataset of 50 patients and an external dataset of 100 patients from Ng Teng Fong General Hospital (NTFGH), Singapore. The model demonstrated excellent performance in classifying stenosis at each cervical disc, achieving near-perfect binary classification (normal/mild versus moderate/severe) agreement for spinal canal ($\kappa = 0.95$) and neural foraminal stenosis ($\kappa = 0.90$). Its sensitivity and specificity ranged from 83.7% to 92.4% and from 87.8% to 98.3%, respectively, across both datasets. In addition, it outperformed radiologists in grading spinal canal ($\kappa = 0.78$ vs. $\kappa = 0.57$ – 0.70) and neural foraminal stenosis ($\kappa = 0.80$ vs. $\kappa = 0.63$ – 0.69), with high recall for cord signal abnormalities (92.3%). These results highlight the potential of the cervical spine model to improve the efficiency and consistency of reporting. The model is now being assessed in a study where radiologists perform stenosis assessment with and without DL assistance [14, 19]. This initial work on a deep

3A

Spine AI report

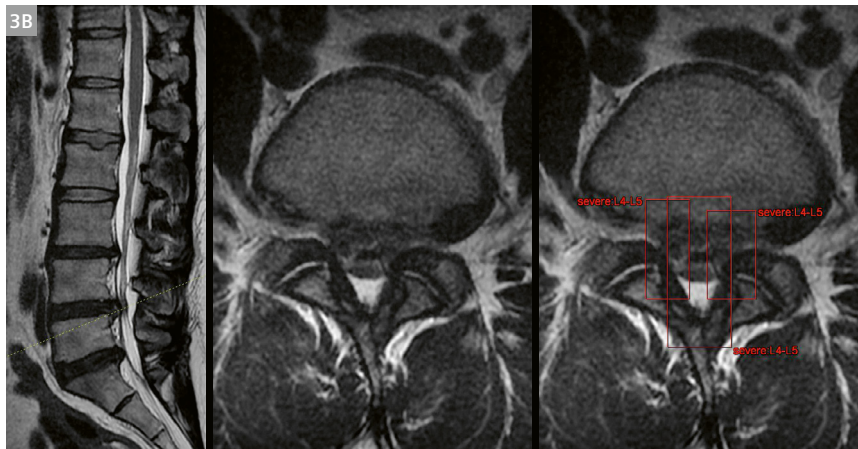
At T12-L1 there is no spinal canal or lateral recess stenosis. There is no neural foraminal stenosis.
 At L1-L2 there is mild spinal canal, no bilateral recess stenosis. There is no neural foraminal stenosis.
 At L2-L3 there is mild spinal canal, no bilateral recess stenosis. There is no neural foraminal stenosis.
 At L3-L4 there is severe spinal canal, moderate bilateral recess stenosis. There is no neural foraminal stenosis.
 At L4-L5 there is severe spinal canal, severe bilateral recess stenosis. There is no right and mild left neural foraminal stenosis.

At L5-S1 there is moderate spinal canal, severe bilateral recess stenosis. There is moderate right and mild left neural foraminal stenosis.

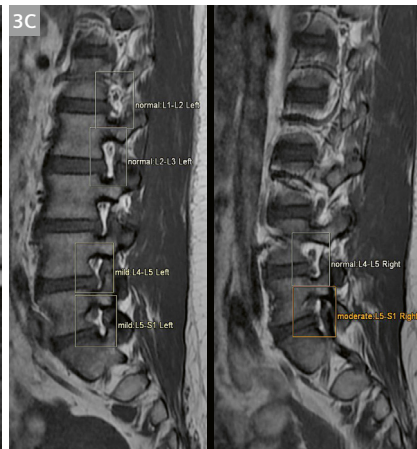
Report based on automated processing!
 Research only, Not for clinical use

Input Data:
 5 SAG T1
 7 AX T2 TOP
 8 AX T2 MID
 9 AX T2 BOT

3B



3C



3 Spine AI predictions for Patient 2, a 36-year-old woman with back pain and suspected lumbar spinal stenosis. An autogenerated Spine AI report is shown in Figure (3A). In Figure (3B), a sagittal T2-weighted image is shown for orientation with a cross-reference line at L4–L5. Stenosis predictions are provided, with severe stenoses at the spinal canal and lateral recesses, mainly due to a large disc protrusion. Figure (3C) shows sagittal T1-weighted images with bounding boxes and stenosis gradings along the bilateral neural foramina, with a nice example of moderate right neural foraminal stenosis at L5–S1.

learning model for MRI cervical spine assessment won a best abstract award at the Royal College of Radiologists (RCR) and NHS Global AI Conference 2025.

Perhaps the most exciting future work is uniting cervical and lumbar models with new thoracic-level algorithms to detect stenoses and spinal cord compression, yielding a unified, whole-spine solution². Such an approach would integrate degenerative pathology detection with oncological assessments for metastatic spinal disease, resulting in a single DL pipeline that offers comprehensive coverage from the cervical to the lumbar region. This is a key aim within the NMRC Clinical Innovator Award, and the team believes it could deliver a greater impact that extends beyond the current lumbar spinal stenosis model [19].

An additional frontier in our team's clinical innovation lies in exploring large language models (LLMs) within radiology. LLMs have great potential to significantly reduce administrative burdens for radiologists by automating tasks such as vetting imaging requests, scheduling appointments and optimizing reporting [20, 21]. Combined with existing clinical data in electronic medical records, prior imaging findings, and AI-generated assessments (e.g., stenosis gradings from the Spine AI model), LLM-based systems

could create more comprehensive and context-aware radiology reports. Integrating image-based and language-based AI would make it possible to further boost productivity, ensure more consistent reporting, and ultimately improve patient care with outputs that are both clinically relevant and seamlessly integrated into the radiology workflow [22].

Conclusion

The journey from developing an AI-based standalone solution for lumbar spinal stenosis to seamlessly integrating it into the clinical workflow illustrates how research, clinical expertise, and industrial collaboration can align to transform clinical care. The DL model created at NUH and NUS has already shown benefits in boosting radiologist productivity, cutting reporting times, and improving the consistency of stenosis gradings. However, these achievements are just the start. As the model evolves and gains more features – such as detecting additional spinal pathologies, analyzing the cervical spine, and identifying metastatic lesions – AI-based tools are poised to become indispensable in everyday radiology. The successful deployment of the Spine AI model as a research application on

4A

Spine AI report

At T12-L1 there is no spinal canal or lateral recess stenosis. There is no neural foraminal stenosis.
At L1-L2 there is no spinal canal or lateral recess stenosis. There is no neural foraminal stenosis.
At L2-L3 there is no spinal canal, no right and mild left lateral recess stenosis. There is no right and left neural foraminal stenosis.
At L3-L4 there is no spinal canal, mild right and no left lateral recess stenosis. There is mild bilateral neural foraminal stenosis.
At L4-L5 there is mild spinal canal, moderate right and mild left lateral recess stenosis. There is mild bilateral neural foraminal stenosis.

At L5-S1 there is mild spinal canal, no bilateral recess stenosis. There is mild right and no left neural foraminal stenosis.

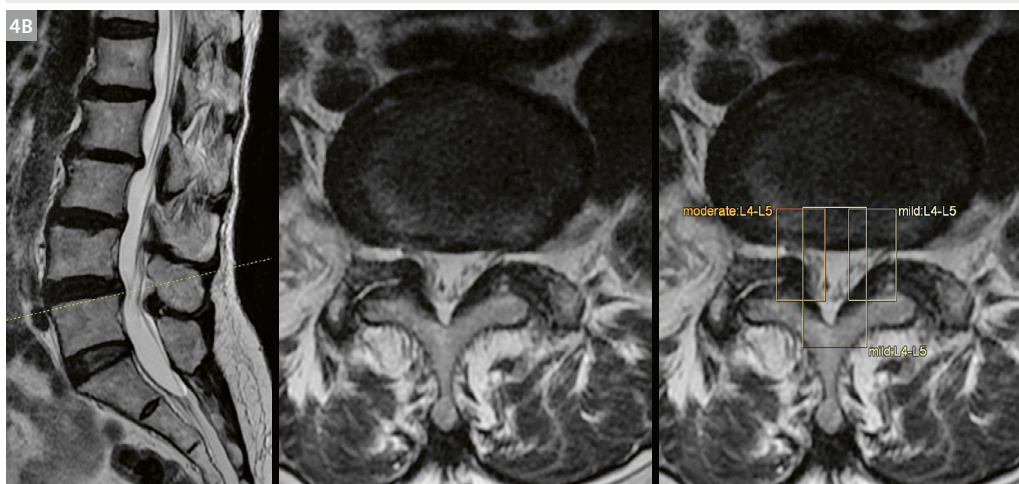
Report based on automated processing!
Research only, Not for clinical use

Input Data:
5 SAGT1
7 AX T2 TOP
8 AX T2 MID
9 AX T2 BOT

4

Spine AI predictions for Patient 3, a 66-year-old woman with back pain and bilateral L5 radiculopathy with right L5 weakness. An autogenerated Spine AI report is shown in Figure (4A). In Figure (4B), a sagittal T2-weighted image is shown for orientation with a cross-reference line at L4–L5. Stenosis predictions are provided, with a nice example of moderate lateral recess stenosis on the right side at L4–5 due to facet joint arthropathy and ligamentum flavum thickening. This impinges on the traversing right L5 nerve root in the lateral recess, compatible with the clinical history.

4B



syngo.via Frontier, backed by a technology development agreement, highlights the practical application of academic AI research in everyday clinical practice and paves the way for further interdisciplinary efforts to improve patient care.

Acknowledgments

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MRI of the Prostate: The Promise of Ultra-Strong Gradients and Advanced Microstructural Imaging

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The need for improved diagnostic tools for prostate cancer

Prostate cancer is one of the most commonly diagnosed cancers worldwide [1, 2], with incidence rates influenced by screening practices, genetic predisposition, and environmental factors [3]. A recent projection estimates a 50% increase in the number of new cases by 2040 [4]. Detection typically begins with a prostate-specific antigen (PSA) blood test and a digital rectal examination, followed by multiparametric MRI (mp-MRI). The gathered information is used to assess the likelihood of a “clinically significant disease,” i.e., identification of cases that are likely to be life threatening and demand further testing or even medical intervention. According to PI-RADS [5] – the standard for determining clinical significance based on mp-MRI – imaging should include standard morphological T1- and T2-weighted, diffusion-weighted imaging (DWI) and dynamic contrast-enhanced (DCE) imaging. Briefly, the DWI exploits the interaction between the random motion of water molecules and the “obstacles” imposed by tissue on the micrometer scale, such as cell membranes, whose characteristics are imprinted in the measured signal. For prostate cancer, DWI is the primary contrast for assessment of lesions in the peripheral zone of the gland. Over the past three decades, DWI has become widely recognized for its role in prostate cancer detection and diagnosis [2, 3] with applications in staging, grading, and treatment planning and monitoring [6].

The apparent diffusion coefficient (ADC) [6] – derived from DWI data with b-values below $1 \text{ ms}/\mu\text{m}^2$ ($= 1000 \text{ s}/\text{mm}^2$) – has been utilized in mp-MRI as a probe of microstructural changes associated with cancer [7]. Its primary strength is that it can identify forms of clinically significant prostate cancer, agreed upon as a Gleason score $\geq 4+3$ [8], with a sensitivity of 93% at patient level [9, 10]. Nevertheless, mp-MRI fails to detect approximately half of the individual tumor foci [11, 12], tends to underestimate the size of large lesions [13, 14], and has poor per-lesion

specificity (41%) [8]. These issues are all likely to be the result of high heterogeneity across cancerous lesions [15]. Moreover, poor geometric accuracy leads to inaccurate delineations of lesions, which limits the use of mp-MRI in radiotherapy treatment planning [16, 17]. Finally, confirmatory biopsy is the gold standard for diagnosis, but it carries risks of infection, bleeding, and sampling errors that may lead to an underestimation of the disease extent. Taken together, this evidence highlights the shortcomings of the existing diagnostic pathway and motivates continued MRI research to improve the specificity and sensitivity of non-invasive diagnostic strategies to minimize the burden and risks associated with invasive testing. However, the information about tissue microstructure that is accurately attainable depends on many factors, including experimental design and signal characteristics, both of which hinge on the performance of the MRI hardware.

A more comprehensive characterization of tumors can be performed by applying stronger diffusion encodings (enabling higher b-values) and coupling the diffusion-weighted signal to a biophysical model or signal representation. An example is the method known as “vascular, extracellular, and restricted diffusion for cytometry in tumors” (VERDICT) [18]. Briefly, VERDICT is a biophysical model describing the tissue as three compartmental water pools: intracellular (diffusion within spherical cells), extracellular (diffusion in stroma and lumen, outside cells/vessels), and vascular tissue (pseudo-diffusion within blood vessels/tubules). It relies on measurements at varying b-values (up to $3 \text{ ms}/\mu\text{m}^2$), diffusion encoding directions, as well as gradient separation and duration (commonly denoted Δ and δ). Recently, the evaluation of VERDICT imaging in a clinical trial with prostate cancer patients (the INNOVATE study) showed that many biopsies that mp-MRI identified as being required could have been avoided, as VERDICT indicated they were not necessary [19]. While emerging “microstructure-MRI” methods like VERDICT offer insights into tumor composition and microstructure, the

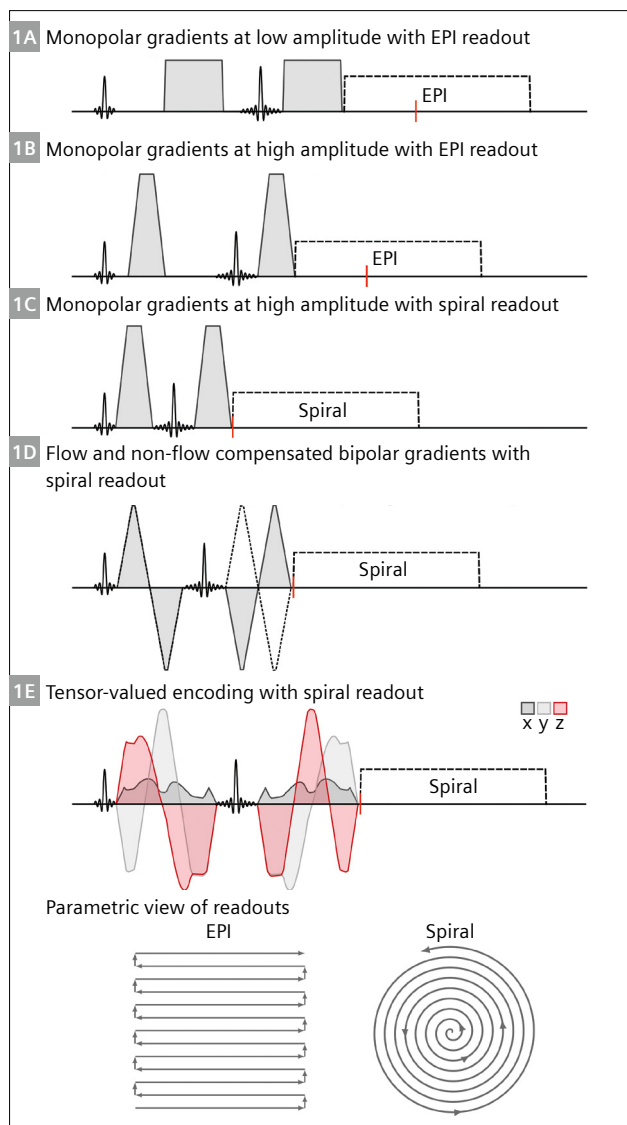
translation of these advances to clinical use is still rather limited. Encouragingly, rapid progress in MRI hardware development is making stronger diffusion encoding more accessible, while improvements in acquisition and optimization techniques are enhancing robustness and efficiency. At the same time, growing clinical expertise and streamlined workflows are helping to bridge the gap

between research and clinical routine. The following sections highlight key technological advancements that are driving the adoption of microstructure MRI in prostate imaging.

Diffusion MRI with strong gradients: New opportunities

Variations in the microstructural properties of living tissue, such as cellularity and tissue organization, create a rich contrast in DWI, providing opportunities to refine methods for more precise, non-invasive tumor characterization. Leveraging these contrasts requires acquisitions that go beyond standard clinical diffusion sequences. The work-horse diffusion-weighted spin echo experiment, i.e., linear b-tensor encoding with monopolar trapezoidal gradients and echo planar imaging (EPI) readout [20–22] (Fig. 1A), has been extensively applied in clinical practice [23, 24]. It is a practical and efficient approach for DWI, with excellent acquisition speed and compatibility across different systems within clinical settings.

Executing unconventional experiments on clinical hardware, with gradient strengths of 40–80 mT/m, incurs relatively long echo times (TE) and reduced signal-to-noise ratio (SNR) per unit time, which can limit the diagnostic utility of the data. Fortunately, the promise of microstructure MRI has motivated several MRI vendors to design systems with much stronger gradients to harness their benefits in research and clinical settings. These MRI systems enable an increase in the accessible measurement space. In certain instances, this results in improvements to measurements that can already be achieved at lower gradient amplitudes, while in other cases, it enables new measurements that would be impractical at lower gradient amplitudes. Specifically, the boosted gradient performance in MRI systems such as MAGNETOM Connectom 1.0 and 2.0¹ (300 mT/m and 500 mT/m), MAGNETOM Cima.X (200 mT/m), and others [25–30] facilitates shorter diffusion encoding times or higher b-values, which improves the tradeoff between b-value vs. TE and SNR (Fig. 1B) [31]. This performance boost is even more apparent in applications that rely on diffusion encoding gradient waveforms with low efficiency, i.e., low b-value per unit time, such as oscillating gradients used to reach short diffusion times or multidimensional arbitrary gradient waveforms for tensor-valued diffusion encoding [32]. By combining these methods, the interrogation of tissue becomes more specific, enabling investigations of subtle features of the microstructure [33, 34]. As a result, DWI with strong gradients has greatly advanced microstructure MRI [35, 36], aiding both research and clinical studies [37–42].



1 Schematic diffusion-weighted MRI spin echo pulse sequences with a parametric view of the readouts. Echo planar imaging (EPI; here with partial Fourier) or spiral with (1A–1C) monopolar gradients, (1D) bipolar flow/non-flow-compensated gradients, and (1E) free gradient waveforms for tensor-valued diffusion encoding. The red line marks achievable echo time (TE). The strong gradients allow for shortening of TE for the same b-value (1B), while alternatives to EPI readouts, such as spiral readouts, minimize the TE even further (1C). The availability of high-performance hardware and efficient readouts allows for the design of exotic diffusion waveforms (1D, 1E), enabling new information to be probed in the diffusion MRI experiment.

¹MAGNETOM Connectom is ongoing research. All data shown are acquired using a non-commercial system under institutional review board permission. Siemens Healthineers does not intend to commercialize the system.

Image readout and reconstruction strategies

The encoding of MR images can be improved (e.g., shorter TE and higher SNR) by more efficient readout strategies, such as spirals [43] or radial [44]. Just like EPI, such trajectories traverse the k -space to acquire image information in the corresponding Fourier space. However, spirals allow for a more tailored readout trajectory, which enables several benefits over EPI. Most notably, spiral trajectories can start close to the TE yet still sample the full k -space, thereby maximizing the time available for diffusion encoding [45] and reducing the TE. Spirals also employ smooth and continuous gradient waveforms, which enables variable sampling density with more efficient signal sampling and superior SNR [46, 47] (Fig. 1C). Therefore, a combination of strong gradients and more efficient image readouts is an excellent match for advanced diffusion encoding at high b -values (Fig. 1D, 1E), such as flow-compensated [48, 49] or tensor-valued encoding [50, 51].

A drawback of spiral readout methods is that the actual k -space trajectory often deviates from the desired, which leads to pronounced artifacts in the image reconstruction [52]. If ignored, such deviations can cause errors in localization or misrepresentation of tissues, ultimately affecting the accuracy of tissue delineation and characterization. The field perturbations that cause this error originate from the object (e.g., motion or tissue susceptibility) and the MR system (e.g., eddy currents [53], gradient system imperfection etc. [54–56]). It is possible to account for such perturbations by using field monitoring [57] and the so-called “expanded encoding model reconstruction” [58]. Doing so means that the actual k -space trajectory can be

employed in image reconstruction, which mitigates artifacts to a high degree of accuracy [45, 59].

Combining ultra-strong gradients and efficient imaging strategies

The combination of ultra-strong gradients and efficient imaging strategies has primarily been used for imaging the brain, but several recent works have focused on imaging below the neck [60–62]. The results from our recent work on a 3T MAGNETOM Connectom system equipped with a 300 mT/m gradient coil (Siemens Healthineers, Erlangen, Germany), in a cohort of men, clearly showcase the added value of strong gradients in prostate imaging. Using DWI protocols with varying gradient amplitude (protocols A–C in Table 1; Fig. 2A–2C), we showed significant improvement in contrast between lesions and healthy tissue when using the protocol at the shortest TE/highest gradient amplitude (Protocol C in Table 1; Fig. 2C). Combining strong diffusion encoding with the expanded encoding model reconstruction [63] further enhanced the prostate cancer lesion delineation compared to the standard acquisition protocol (Protocols D–F in Table 1; Fig. 2D–2F). The high b -value images, free from artifacts, demonstrated exceptional anatomical fidelity, with sharp tissue boundaries and improved alignment across shots, even at the highest b -values of 3 ms/ μm^2 [63].

We compared the SNR gains afforded by ultra-strong gradients (~300 mT/m) to clinical-like gradient amplitudes (~40–80 mT/m) when imaging the prostates of a cohort of healthy controls and patients [60]. It was shown that

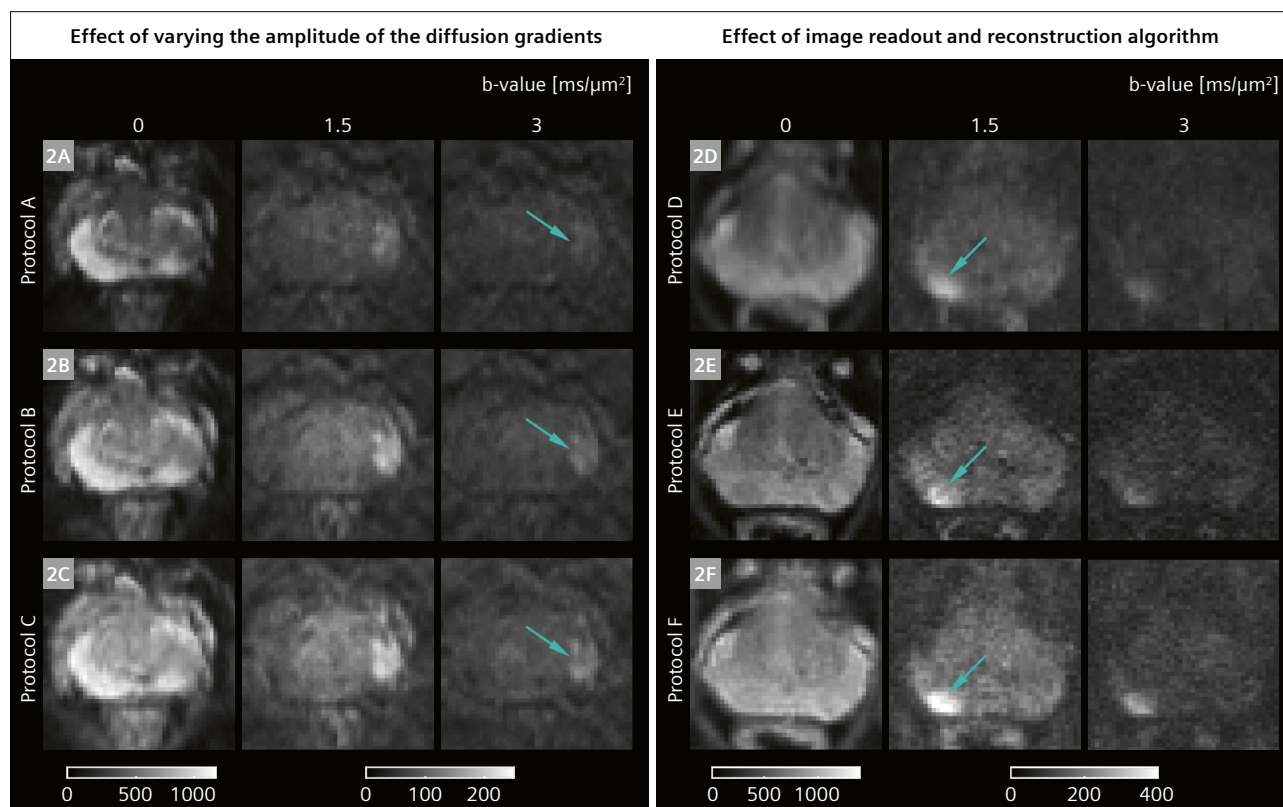
Identifier	Acquisition	b-values [ms/μm ²]	Gradient amplitude [mT/m] / Slew rate [T/m/s]	Diffusion encoding duration/separation [ms]	Readout type	TE [ms]	Protocol duration min:sec
A	Monopolar pulsed gradients	0.05, 0.5, 1.5, 2, 3	40/200	26/48	EPI	95	6:35
B			80/200	16/32		70	
C			273/111	5/25		54	
D			247/83	5.7/23.3		53	3:30
E							
F					Spiral	35	
G	Bipolar gradients	0.01, 0.02, 0.03, 0.05, 0.1, 0.15, 0.2, 0.3, 0.5	220/65	14.5/22.6		41	7:00
H	Tensor-valued encoding (linear and spherical b-tensor)	0.05, 0.2, 0.5, 0.8, 1.1, 1.5, 2, 3	220/< 60	21.6/29.9		54.8	9:50

Table 1: A summary overview of key protocol parameters from our studies, which are showcased in this manuscript. Importantly, the resolution and repetition time were kept similar across protocols. Note that the protocol identifiers are independent from the labels used in Figure 1.

employing 300 mT/m gradients yields statistically significant gains in SNR, particularly evident at 1.5 ms/μm², for which 15% and 35% gains in comparison to protocols with 80 mT/m and 40 mT/m, respectively, were observed (Fig. 3A). Furthermore, spiral readouts provided an additional SNR gain of 30%–45%, depending on the b-value [63]. The contrast-to-noise ratio (CNR) of the cancerous to healthy tissue was highest for the protocol using 300 mT/m across all b-values, while within each protocol, the CNR was the highest at b = 1.5 ms/μm² [60]. In the side-by-side comparison of protocols by an experienced radiologist, the protocol with the highest gradient amplitude was generally preferred for lesion delineation.

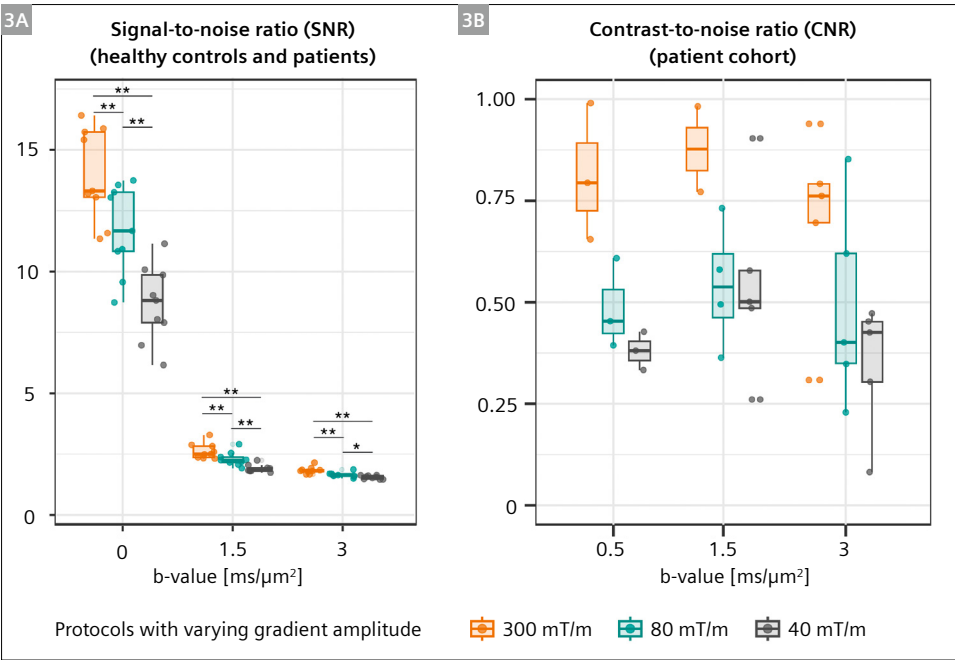
Increasing gradient amplitude to reduce TE introduces challenges related to T2 weighting. While diffusivity itself is an intrinsic property, in heterogeneous tissues, the contribution of each tissue compartment to the diffusion-weighted signal is influenced by both the volume fraction

and the T2-relaxation times of that compartment. As such, the contrast in DWI (and the derived parameters) depends on the TE – with species having shorter T2 values potentially contributing more to the overall signal due to shorter TE (e.g., restricted fraction [64]). The tissue diffusivity maps (Fig. 4A–4C) show differences attributable to differences in diffusion time and relaxation effects across protocols A–C, as outlined in Table 1 [60]. For the protocol with high gradient amplitude and short TE (Protocol C, Fig. 4C), the maps exhibit greater homogeneity compared to protocols A and B, and show lower estimates of diffusivity at the gland level. The highest contrast between cancerous lesions and healthy tissue is observed in Protocol A (e.g., the patient presented in the bottom row of Fig. 4). However, the observed higher diffusivity in healthy tissue and slightly lower diffusivity in the tumor, as seen in data from protocols with low gradient amplitudes, are likely biased due to the lower SNR of the data.

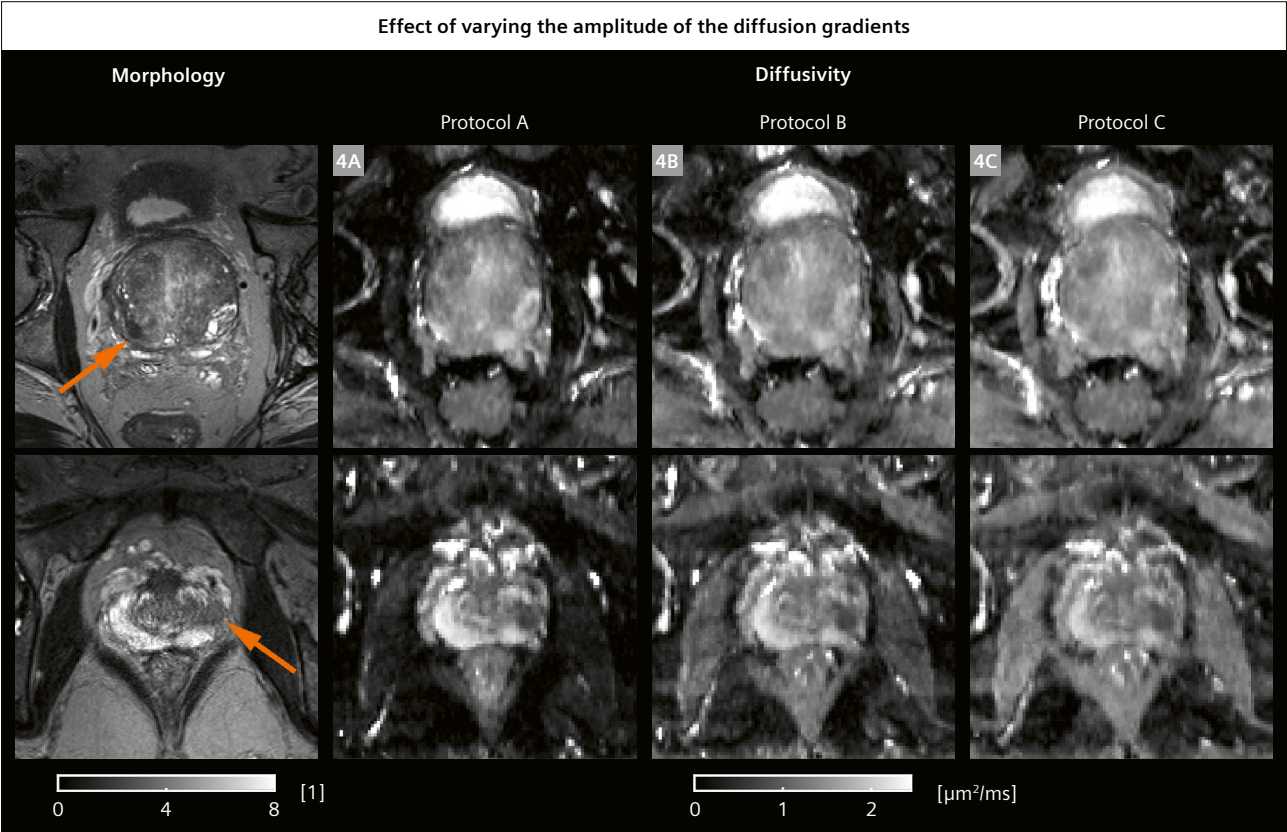


2 Qualitative overview of the diffusion data (direction-averaged) from two patients with prostate cancer, acquired using sequences with different maximum gradient amplitude/readout or reconstructed using different algorithms. **(2A–2C)** Data from a patient (67 years old) with prostate cancer of 3+4 Gleason score acquired with a prototype sequence allowing for arbitrary setting of maximum gradient amplitude ($G_{\max}$) with varying diffusion encoding/separation times and TE parameters (Protocols A–C in Table 1). A visual improvement in contrast between lesions and healthy tissue is observed when using the protocol at the shortest TE (2C). **(2D–2F)** Data from a patient (53 years old) with prostate cancer of 3+3 Gleason score acquired with the product EPI (2D) and prototype sequences² allowing for the use of EPI (2E) or spiral (2F) readouts (protocols D–F in Table 1). Improved geometric consistency – better preserved edges of prostate gland – across different shots is obtained when accounting for field perturbations using advanced image reconstruction (2E, 2F). The data presented in 2A–2D was reconstructed with the default reconstruction program from Siemens Healthineers. Data from [59, 62].

²Work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.



3 (3A) Signal-to-noise ratio (SNR) for selected b-values. The dots represent the estimate of SNR at the subject level (healthy control or patient), and the box plot represents the median and interquartile range of SNR per given protocol at a group level (*p < 0.05, **p < 0.01). (3B) Contrast-to-noise ratio (CNR) for selected high b-values. Dots represent CNR at the subject level (here, patients only), and the boxplot represents the median and interquartile range per given protocol at a group level. The data was published in [59].



4 Morphology and diffusivity maps from the diffusion tensor imaging (DTI) fit in two representative patients with prostate cancer (top row: 68 years old, 3+3 Gleason score; bottom row: 67 years old, 3+4 Gleason score). The estimates (columns 4A–4C) were obtained using data acquired with different maximum gradient strengths (Protocols A–C in Table 1) and show differences presumably due to different diffusion time and relaxation effect. The best contrast between cancerous lesions and healthy tissue is observed in the protocol using 40 mT/m (Protocol A). However, the diffusivity estimated at longer echo times may also be inaccurately estimated due to poor SNR and noise floor effects. Cancerous lesions are indicated with orange arrows. The data presented was reconstructed with the vendor's default reconstruction program.

Imaging blood perfusion by intravoxel incoherent motion

Intravoxel incoherent motion (IVIM) [6] imaging is a DWI method that is sensitive to the perfusion of capillary blood and provides promising imaging biomarkers in prostate cancer [65]. The method employs conventional diffusion encoding at densely distributed b-values up to $0.5 \text{ ms}/\mu\text{m}^2$. The model used to analyze the data assumes two water pools: one describes pseudo-diffusion of blood in randomly oriented capillaries/blood vessels, and the other describes diffusion of water in tissue. Crucially, the parameter related to vasculature is the key characteristic of interest in oncology as a marker of ongoing hypoxic processes [65, 66]. However, the parameter estimation is ill-posed and has poor repeatability, which impedes clinical adoption [67, 68]. The visual quality of resulting parameter maps can be improved by performing segmented fitting [67] or using Bayesian methods [69], however, if the assumptions of such methods are violated, the result may be biased [68]. Another strategy for improved parameter estimation is to expand the dimensionality of the measurement, thereby injecting independent information that stabilizes the estimation [70]. The addition of “flow-compensated,” or more accurately “velocity-compensated,” diffusion encoding (Fig. 1D) was proposed and demonstrated as a viable solution to these problems [49, 71].

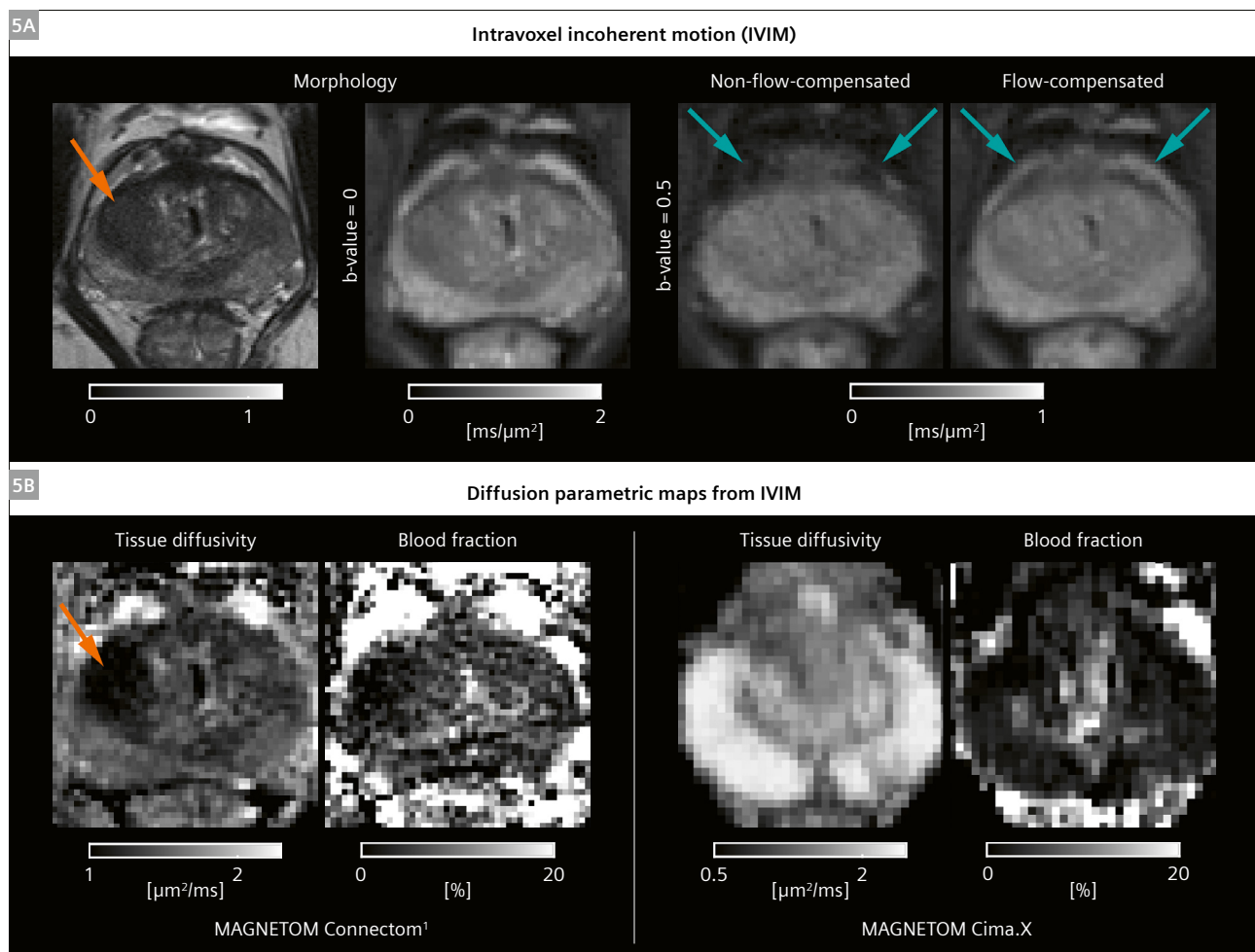
By leveraging the technological advancements (strong gradients and advanced image reconstruction) and deploying a tailored experimental design, we achieved unprecedented data quality and IVIM parametric maps [72, 73] (Protocol G in Table 1). Data acquired with our prototype pulse sequence had markedly high structural fidelity, high SNR, and could distinguish regions where velocity encoding has a major impact on the signal, e.g., the venous plexus, which is a mesh of small blood vessels (Fig. 5A). The signal and parameter maps could be reconstructed without visible distortions (a single-subject example shown in Fig. 5B, left column). Across a cohort of patients, the fitting using only non-flow-compensated acquisitions yields a slightly underestimated tissue diffusivity and notably overestimated blood fraction, while analysis using complete flow/non-flow-compensated data provides values of the superior accuracy as supported by simulations [71, 74]. Across subjects, cancerous lesions were distinguished by markedly lower diffusivity and lower blood fraction than peripheral and transitional zones. Future work will comprise an evaluation of IVIM parameters in a larger cohort to assess its full potential as perfusion marker, e.g., in radiotherapy. Recently, we have translated the IVIM protocol to a 3T MAGNETOM Cima.X scanner (Siemens Healthineers, Erlangen, Germany) [75], obtaining similar values of the diffusion encoding duration/separation as in the protocol on the MAGNETOM Connectom 1.0 scanner. Our pilot

analysis from healthy volunteers (a single-subject example shown in Fig. 5B, right column) provided an estimate of blood fraction in line with our previous experiments.

Imaging heterogeneity

Another promising extension to conventional DWI is tensor-valued diffusion encoding [76–79]. It has gained recognition for its ability to distinguish between macroscopic and microscopic diffusion anisotropy [76, 80–82]. The method relies on two methodological ingredients: diffusion encoding at moderate-to-high b-values ($2\text{--}3 \text{ ms}/\mu\text{m}^2$) and encoding with b-tensors with varying “shapes” [51]. The shape feature arises from diffusion encoding along more than one direction per shot, where the shape describes whether the encoding was linear (along one direction as in conventional diffusion MRI), planar (equal encoding in a 2D plane), spherical (equal encoding in all directions), or anywhere in between [83]. An example of spherical b-tensor encoding is shown in Fig. 1E. Such measurements enable the exploration of three effects that are otherwise entangled: microscopic diffusion anisotropy (sub-voxel diffusion anisotropy), isotropic heterogeneity (sub-voxel variation of diffusivity), and orientation coherence (sub-voxel directional order of anisotropic diffusion) [80]. A recent study demonstrated that such an approach can map different sources of heterogeneity in tumors such as gliomas and meningiomas [84]. In prostate cancer, two preliminary studies have investigated the feasibility of b-tensor encoding in the assessment of tissue heterogeneity using clinical systems [85, 86]. Although the studies demonstrated technical feasibility and provided an initial insight into the microscopic anisotropy of prostate tissue in vivo, they were limited by poor spatial resolution, low SNR, and marked geometric distortions.

In a preliminary study, we have shown the utility of high-fidelity b-tensor encoded DWI data for characterizing the prostate (Protocol H in Table 1, Fig. 6) [87], wherein we addressed the limitations of the previous studies. We show that the peripheral zone exhibits lower anisotropic microscopic kurtosis than the transitional zone, suggesting a higher presence of elongated cells and stroma in the transitional zone. Cancerous lesions in the peripheral zone exhibited elevated isotropic and anisotropic kurtosis compared to healthy tissue, potentially reflecting epithelial changes, such as cell proliferation with varied diffusivities, and/or increased stroma, which raises microscopic kurtosis anisotropy (evidenced by higher microscopic fractional anisotropy). The pulse sequence – based on the diffusion-weighted spin echo product sequence from Siemens Healthineers – is provided by Lund University through the teamplay platform from Siemens Healthineers under the name “Free waveform (FWF) diffusion MRI.”²



5 (5A) Example intravoxel incoherent motion (IVIM) dataset (Protocol G in Table 1) from a patient with prostate cancer (69 years old, 3+3 Gleason score). The orange arrow points to a cancerous lesion and the blue ones depict venous plexus around the prostate gland, which is visibly the most sensitive to the flow and non-flow-compensated diffusion encoding. (5B) IVIM parameter maps from the same case subject as in (5A) scanned on the MAGNETOM Connectom (left), and from a healthy volunteer (63 years old) scanned on the MAGNETOM Cima.X (right). The data acquired on the MAGNETOM Connectom¹ scanner maintains superior quality (spiral readout and “expanded encoding model” reconstruction). The successful translation of the experiment to a strong-gradient clinical MR scanner (single-shot EPI readout and default reconstruction program) yielded parameter maps which are consistent with our previous results. *Data from [71, 92].*

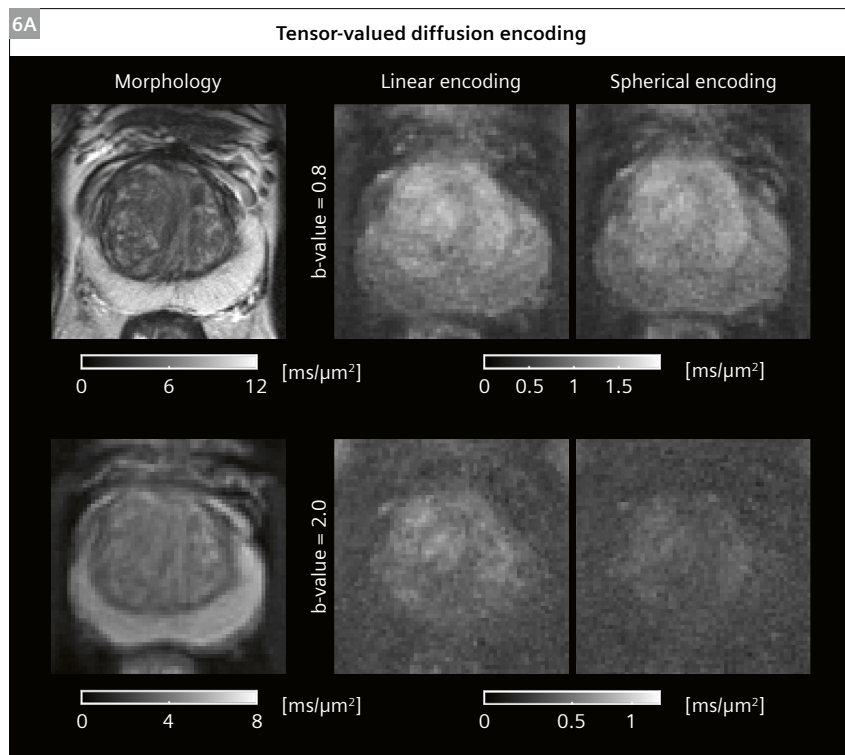
Multidimensional MRI: Diffusion time, relaxation, and beyond

An often-overlooked experimental variable in DWI is the diffusion time, which refers to the duration during which diffusive motion influences the signal amplitude during each acquisition. By increasing the available gradient amplitude, we can explore a wider range of diffusion times for the same b-value, allowing us to probe faster and slower diffusive behaviors. The information from such time-modulated diffusion, called time-dependent DWI, allows us to gauge additional information, which increases the specificity of the measurement in the assessment of microstructural properties of cellular environments [88, 89]. One of the most common examples of such

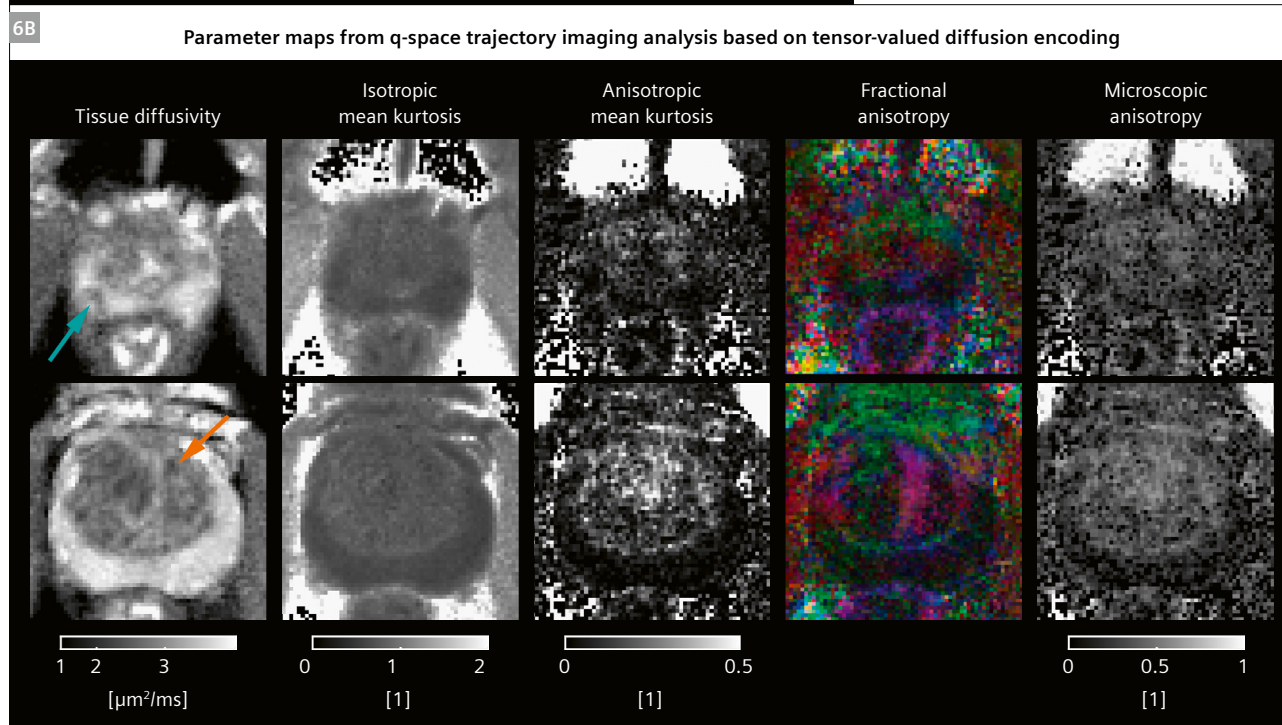
a measurement is the oscillating-gradient spin echo experiment [90], which allows for shorter diffusion times than the conventional pulsed-gradient spin echo experiment. A combination of these conventional and oscillating diffusion encoding waveforms is leveraged in the “imaging microstructural parameters using limited spectrally edited diffusion” (IMPULSED) framework [91]. In brief, the IMPULSED model combines multiple short diffusion time oscillating gradient measurements with a single long diffusion time regime trapezoidal gradient acquisition to quantify the characteristic size of the restricting geometry and the intracellular signal fraction. A recent study [92] reported that time-dependent DWI parameters, especially the cellularity index (defined as the ratio of the intracellular volume fraction and cell diameter), achieved

superior accuracy in discriminating more advanced cancerous disease from clinically insignificant changes. The diffusion time dependence of the signal has not been explicitly accounted for in tensor-valued encoding, until very recently [93, 94]. An extended version of tensor-valued encoding that considers diffusion time jointly

can quantify how the diffusion, microscopic diffusional variance/anisotropy, and their covariances vary with diffusion time. Currently, the method is being tested in animal studies with human prostate cancer xenografts, and in healthy and prostate cancer patients. In line with the literature [95, 96], our results indicate that restrictions



6 **(6A)** Acquired images in a pilot tensor-valued experiment (Protocol H in Table 1) in a prostate cancer patient. Diffusion-weighted images acquired with different encoding b-tensor (different shape and strength of encoding, b-value) show similarities in contrast at low b-value, while the contrasts differ at higher b-values. This showcases the sensitivity of encoding to different properties of underlying microstructure. **(6B)** Example maps from tensor-valued diffusion MRI scan in the human prostate. Two slices with low-grade cancerous lesions in peripheral and transitional zones (arrows) are shown. While diffusivity is decreased in lesions, mean kurtosis maps – isotropic and anisotropic – are increased in peripheral-zone lesion. The effect in parameters in the transitional-zone lesion is not so clear. At macroscopic level, anisotropy is quite low in the transitional zone, but at microscopic level, it is notably higher.



and diffusion time dependence generate the estimated parameters heavily dependent on the experimental parameters [93]. Given the ongoing interest in time-dependent DWI in tumors as a “quantitative imaging biomarker,” this methodology is promising and warrants a systematic validation of the method in a clinical cohort.

DWI can be further enhanced by incorporating joint measurements of relaxation, a strategy demonstrated in studies using phantoms, and in animal and human tissues [97–102]. In prostate imaging, this approach has led to the development of a three-pool model, with methods such as hybrid multidimensional MR imaging (HM-MRI) [103] and relaxed VERDICT (rVERDICT) [104]. These models explicitly account for transverse relaxation effects, improving the accuracy of microstructural characterization. HM-MRI has great potential to improve the diagnosis of prostate cancer and determine its aggressiveness, as fractional volumes of prostatic tissue composition (lumen, stroma, and epithelium), estimated non-invasively and by pathologists, were reported to be similar to the true tissue composition [105]. The deep learning-based rVERDICT allows for accurate, fast, and repeatable analysis, bringing good discriminative power to disease staging. Additionally, the use of two-dimensional correlation experiments has provided further insights into the complex relationship between diffusion and relaxation in prostate tissue [106].

Conclusion: A virtual microscope for the prostate

Extending conventional DWI to include advanced diffusion encoding and image readouts brings the family of methods closer to the promise of a “virtual microscope.” This development hinges on high-performance MRI systems with a special weight placed on the gradient system. Importantly, a vast majority of advanced DWI methods stand to gain marked improvement in data quality from ultra-strong gradients and more efficient readout strategies, and the exploration of biomarkers can be extended to include experimental designs that were previously prohibitively slow or yielded insufficient SNR. These developments are especially exciting for the prostate as they promise high-fidelity high-resolution images that reflect ever more subtle features of tissue microstructure, enhancing detection, diagnostics, and treatment monitoring.

Several research studies have already presented promising springboards for prostate cancer characterization using frameworks beyond clinically adapted conventional diffusion encoding and the quantitative ADC map. Our research provides early insights into what novel information can be gained from new hardware, and drives further development in the context of prostate cancer imaging. With powerful MRI systems becoming commercially available worldwide, new imaging biomarkers will

be facilitated by strong-gradient technology and are likely to play a fundamental role in everyday clinical diagnosis, contributing to the development of individualized precision medicine.

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Fast Abdominal MRI with Deep Learning Reconstruction: Practical Experience from Zhongshan Hospital

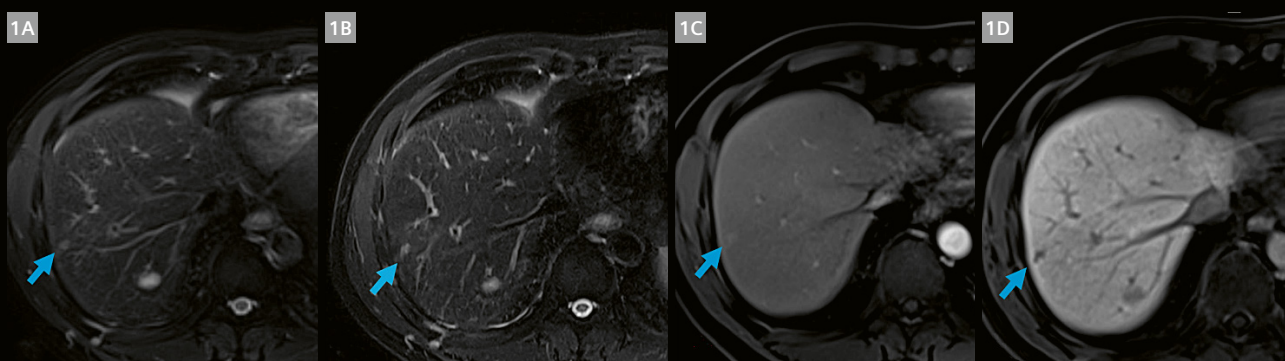
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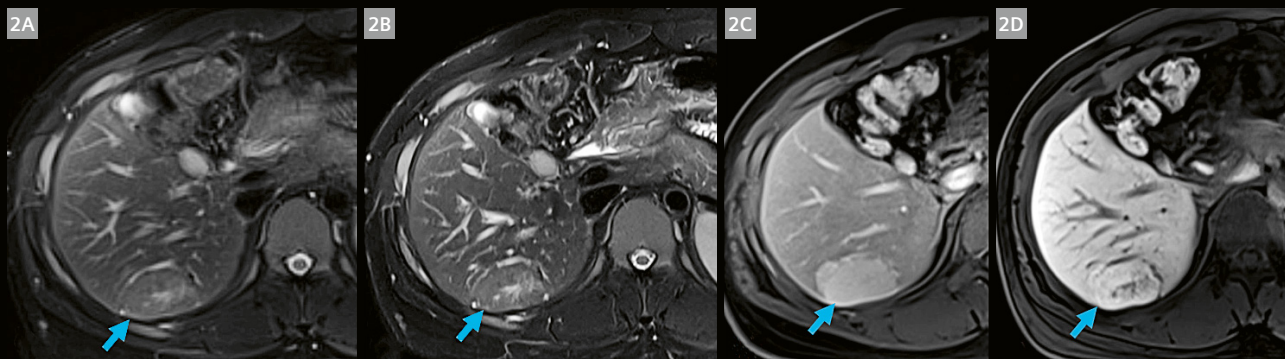
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In recent years, deep learning image reconstruction (DLR) technology has made significant progress in the field of medical imaging, particularly in magnetic resonance imaging (MRI) [1]. Since abdominal MRI is a crucial diagnostic tool for liver, biliary, and pancreatic diseases, its image quality directly impacts diagnostic accuracy. In abdominal MRI, DLR technology can help to significantly reduce

acquisition time while maintaining or enhancing image quality through optimized image acquisition, reconstruction, and denoising [2–4]. This article presents our practical experience in applying DLR technology from Siemens Healthineers to abdominal MRI, including T2-weighted imaging (T2WI), magnetic resonance cholangiopancreatography (MRCP), and diffusion-weighted imaging (DWI).



1 A patient with a subcentimeter hepatocellular carcinoma (HCC) lesion (arrow) in the right lobe of the liver. Compared to the triggered BLADE (1A), single-breath-hold HASTE (1B) shows better conspicuity for the small lesion; T1WI in the arterial phase (1C) and hepatobiliary phase (1D) confirms the presence of the lesion.



2 A patient with focal nodular hyperplasia (FNH) (arrow) in the right lobe of the liver. Compared to the triggered BLADE (2A), single-breath-hold HASTE (2B) shows the lesion more clearly. The lesion also appears on T1WI in the arterial phase (2C) and hepatobiliary phase (2D).

As hepatobiliary diagnosis and treatment constitutes one of our hospital's four major specialties, our radiology department performs around 850 MRI scans daily, including approximately 300 time-consuming liver examinations. Despite having 12 MRI scanners for routine clinical use, improving examination efficiency and reducing patient waiting times remain urgent priorities. Therefore, implementing DLR technology to enhance overall MRI throughput – particularly for abdominal scans – while maintaining or even improving image quality holds significant appeal for our institution.

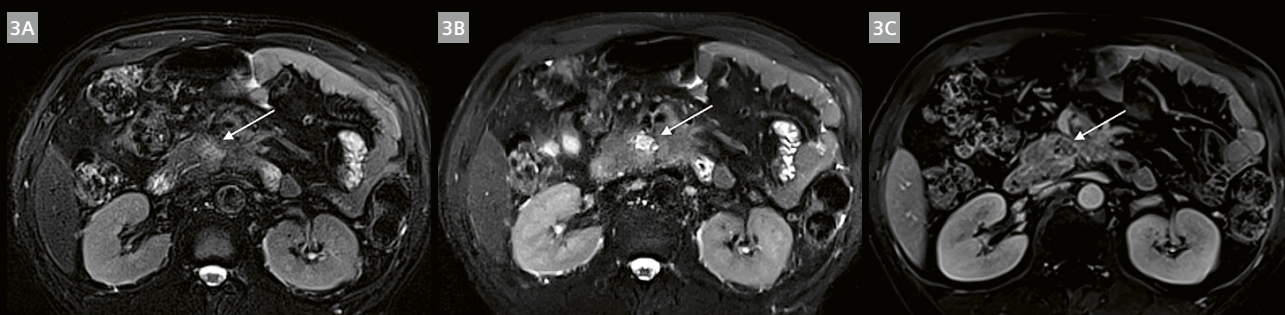
Application of DLR in T2WI and MRCP

In abdominal MRI, T2WI is a fundamental sequence, playing a vital role in visualizing anatomical structures and detecting abdominal pathologies. Conventional T2WI typically employs respiratory-triggered or multi-breath-hold TSE/BLADE sequences, which are prone to respiratory pattern interference (prolonged acquisition time, inconsistent image quality, slice discontinuity) or patient discomfort from repeated breath-holding.

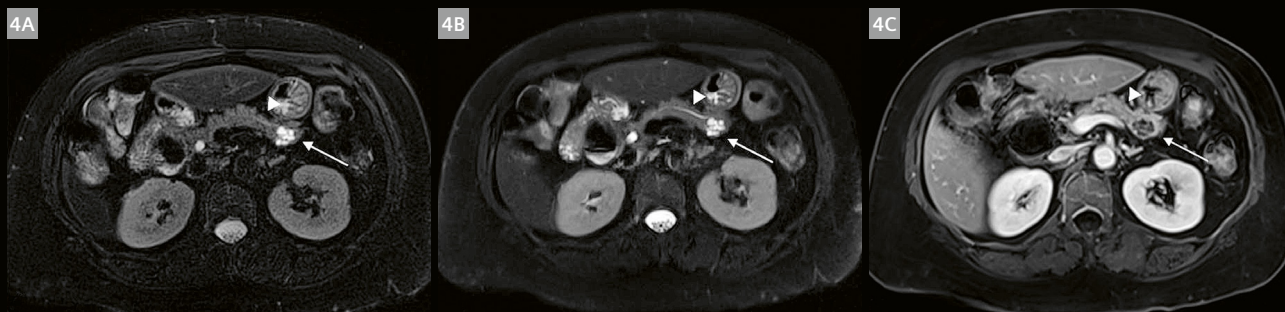
Using DLR-accelerated HASTE (DL HASTE) sequences on our BioMatrix 3T MRI scanner (Siemens Healthineers, Erlangen, Germany), we achieved single-breath-hold T2WI with a 16-second breath-hold. Compared with conventional BLADE sequences, this approach significantly improves scanning efficiency while maintaining stable, high-quality imaging. Figures 1 and 2 demonstrate the comparison between single-breath-hold DL HASTE (16 seconds) and conventional triggered BLADE (around 3 minutes) for T2W liver imaging.

Pancreatic MRI normally demands higher image quality than MRI of other abdominal organs, due to the gland's thin anatomical profile and the typically small size of pancreatic lesions. We applied single-breath-hold DL HASTE for high-resolution thin-section pancreatic T2WI, achieving enhanced lesion conspicuity with reduced respiratory motion artifacts [5]. Figures 3 and 4 compare single-breath-hold DL HASTE (16 seconds) with conventional 4-breath-hold BLADE (102 seconds) for pancreatic T2WI.

MRCP is a critical technique for evaluating biliary and pancreatic duct pathologies. However, conventional MRCP acquisition suffers from prolonged scan times and



3 A 66-year-old male patient with a pancreatic head tumor. The pancreas and pancreatic lesion (arrow) can be clearly seen in DL HASTE (3B). However, the shape of the lesion depicted in the BLADE image (3A) appears indistinct and lacks clear definition. The contrast-enhanced T1WI (3C) confirmed the pancreatic head tumor (arrow) [5].



4 A 71-year-old female patient with cystic lesions of the tail of the pancreas. The pancreas and pancreatic lesion (arrow) can be clearly seen in both DL HASTE (4B) and BLADE (4A); The pancreatic duct (triangular arrow) is clearly visualized on DL HASTE, whereas conventional BLADE provides poor ductal delineation and lower SNR. The contrast-enhanced T1WI (4C) shows the pancreatic lesion (arrow) [5].

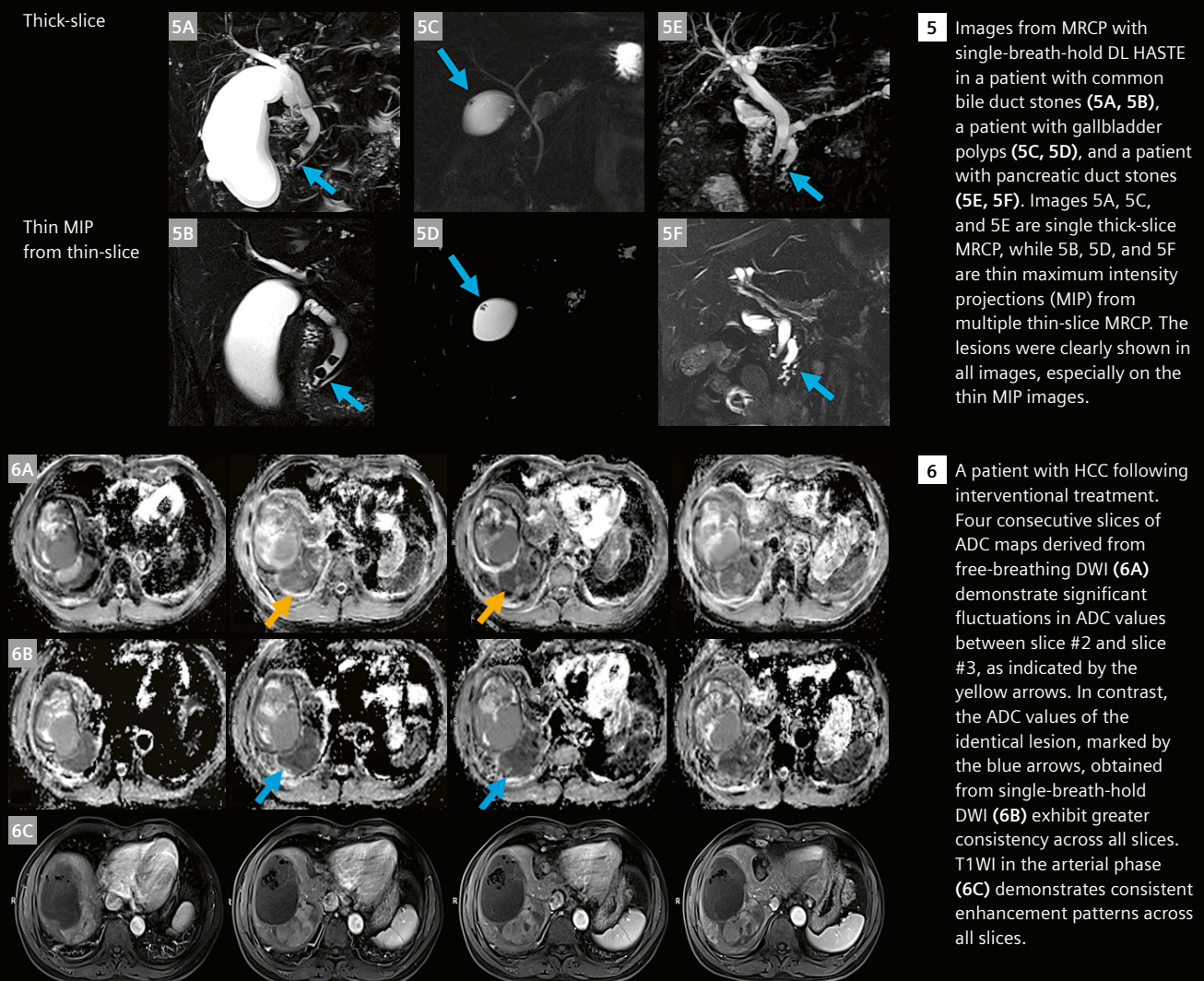
sensitivity to motion artifacts. We implemented single-breath-hold MRCP using DL HASTE with a duration of 9 seconds for thick-slice protocols, and 19 seconds for thin-slice protocols. This substantially improved scan success rates while maintaining satisfactory performance in duct visualization and lesion detection. Figure 5 shows three clinical cases that exemplify the excellent ability of this sequence to display biliary/pancreatic duct structures and enable identification of pathological findings.

Application of DLR in DWI

DWI provides insights into tissue microstructure by detecting water molecule diffusion. It is a critical functional imaging technique for evaluating abdominal pathologies [6, 7]. The apparent diffusion coefficient (ADC) derived from DWI plays a significant role in diagnosing abdominal

lesions by quantifying water molecule diffusion within tissues. However, traditional DWI acquisitions – whether using respiratory triggering, multiple breath-holds, or free breathing – are prone to slice misregistration between different b-values due to respiratory motion. This leads to inaccuracies in ADC calculations, where the same lesion may exhibit drastic ADC variations across adjacent slices, a phenomenon termed the “yin-yang sign.” By using DLR, we achieved a single-breath-hold duration of approximately 20 seconds for abdominal DWI with two b-values, thereby minimizing positional mismatches between b-values and improving the accuracy of ADC estimation and diagnostic reliability.

Figure 6 compares ADC maps from free-breathing DWI (1:26 minute) and single-breath-hold DWI (both with $b = 50$ and $b = 800 \text{ s/mm}^2$).

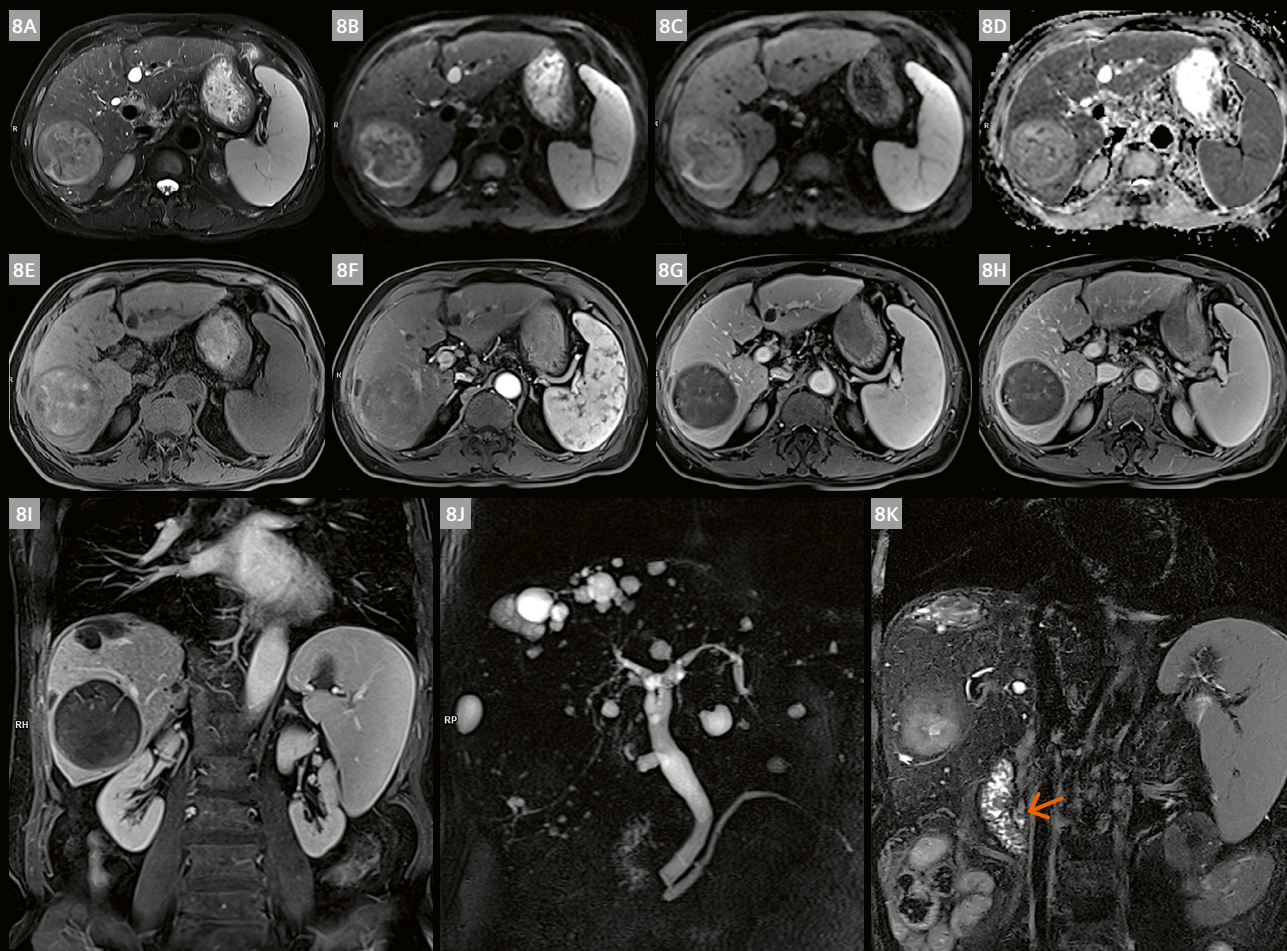


Fast abdominal MRI protocol

We have now integrated single-breath-hold sequences including DL T2WI, DL MRCP, and DL DWI into the routine abdominal MRI protocol on our BioMatrix 3T systems, establishing a fast abdominal MRI workflow. The total image acquisition time for this protocol is 3 minutes and 47 seconds, and the whole workflow takes roughly 10 minutes, including needle insertion and contrast agent administration. Figure 7 illustrates the entire protocol, while Figure 8 presents images from a post-treatment HCC patient, acquired with this protocol.

Abdomen			ISO					
1	localizer							00:12
2	t2_haste_fs_tra_p4_352_bh							00:16
3	t1_vibe_dixon_tra_p4_bh_352							00:14
4	ep2d_diff_b50_800_tra_p2							00:20
5	Prepare injector.							
6	Care Bolus							01:22
7	t1_vibe_fs_tra_p4_384_artery							00:14
8	t1_vibe_fs_tra_p4_384_Veinous							00:14
9	t1_vibe_fs_cor_p4_bh_352							00:13
10	t2_haste_fs_cor_p2_thick_384...							00:09
11	t2_haste_fs_cor_p4_thin_352...							00:19
12	t1_vibe_fs_tra_p4_384_delayed							00:14
								Σ 03:47

7 Fast abdominal MRI protocol integrated with single-breath-hold sequences including DL HASTE for T2WI and MRCP, and DL DWI.



8 Images of a post-treatment HCC patient obtained using the fast abdominal MRI protocol. The image quality is excellent, with lesions clearly visualized in all sequences. (8A) T2WI, (8B) DWI b50, (8C) DWI b800, (8D) ADC, (8E) pre-contrast T1WI, (8F) arterial-phase T1WI, (8G) venous-phase T1WI, (8H) delayed-phase T1WI, (8I) coronal delayed-phase T1WI, (8J) single thick-slice MRCP, (8K) multiple thin-slice MRCP.

Conclusion

The application of DLR technology in abdominal MRI has markedly reduced acquisition time while improving image quality and diagnostic accuracy. Shortening scan duration not only enhances scanner throughput and reduces operational costs, but also improves patient comfort and minimizes motion artifacts. We anticipate broader clinical adoption of this method to elevate the overall efficiency of MRI.

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Body Imaging at 7T: A Path Toward Clinical Translation

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Introduction to body imaging at ultra-high fields

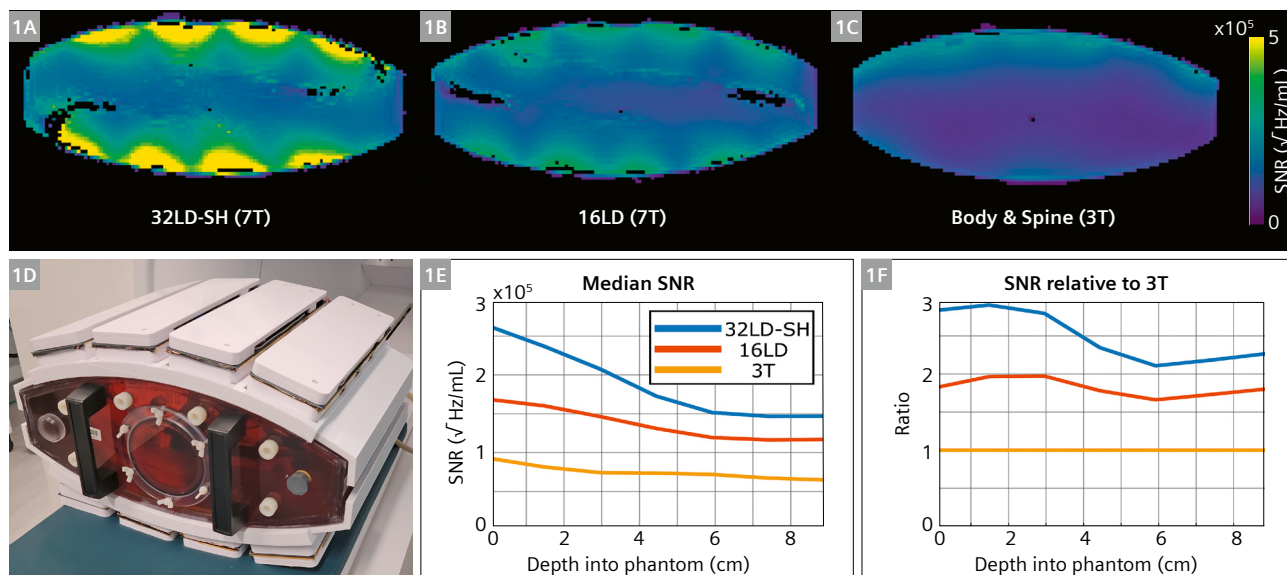
The evolution of ultra-high-field (UHF) magnetic resonance imaging (MRI) at 7 Tesla (7T) is gradually unlocking opportunities for high-resolution anatomical and functional imaging of the human torso¹. Over nearly two decades, progress in radio-frequency (RF) hardware, transmit chain architectures, and electromagnetic field management strategies have addressed many of the challenges associated with transmit B1 (B1+) inhomogeneity and specific absorption rate (SAR) constraints. Key developments include the transition from single-channel to parallel-transmit (pTx) systems, the development of multichannel transceiver arrays, and advanced RF shimming and pulse design techniques that balance B1+ efficiency and homogeneity. High-quality images obtained at 7T and 10.5T in the pelvis, abdomen, and chest demonstrates the poten-

tial for clinical translation, while ongoing innovations in real-time SAR monitoring, calibration methods, and universal solutions promise to streamline workflows. Despite regulatory and technical hurdles, 7T MRI is poised to transform precision medicine by merging high sensitivity with molecular insights. While UHF body imaging is reviewed in detail in other published work [1–4 and references therein], in this article, we will review the interplay between technological innovation and anatomy-specific imaging at our center, concluding with a roadmap for clinical adoption.

Theoretical foundations of 7T body MRI

Benefits of body imaging at 7T

UHF body imaging holds significant promise for improved performance in multiple anatomical targets throughout



1 Receive performance of the 32-channel shielded loop-dipole array (32LD-SH) [18]. The SNR maps compare **(1A)** the new 32LD-SH with **(1B)** our previous 16-channel transceiver loop-dipole array (16LD) and **(1C)** the typical body imaging setup on 3T consisting of an 18-channel body array anteriorly with a 32-channel spine array posteriorly. **(1D)** shows the physical coil positioned on the phantom for SNR validation and testing. The SNR performance is shown for each coil over increasing depths given as both **(1E)** the calculated SNR and **(1F)** the ratio with respect to the 3T setup. The holes in the 7T datasets are where B1+ was not sufficiently defined to provide accurate SNR measurements.

¹Abdominal imaging at ultra-high-field MRI is a research application and is not released for clinical use. It is not for sale in the U.S. and in other countries and its future availability cannot be ensured.

the torso in both research and clinical settings [1–4]. Advantages include increased signal-to-noise ratio (SNR), enhanced parallel imaging performance, and favorable relaxation properties. As in the brain, the presence of increased susceptibility contrast provides improved sensitivity to oxygenation and exogenous contrast media [5]. Prolonged T1s lead to improvements in a host of applications including in-flow angiography and arterial spin labeling [6, 7]. To realize these advantages and achieve wider adoption, several technical developments are required, including improvements in RF coils and strategies to manage the electromagnetic fields.

Electromagnetic challenges at ultra-high fields

At 7T, the Larmor frequency of protons is 300 MHz, which results in RF wavelengths (~11 cm in biological tissues) shorter than torso dimensions, leading to constructive/destructive interference patterns and severe B1+ inhomogeneity. This phenomenon creates spatially varying flip angles that compromise image uniformity both in terms of signal intensity and contrast. Imaging in the torso necessitates more advanced RF management strategies, including static and dynamic pTx methods [8]. The inverse relationship between transmit efficiency ($B1+/\sqrt{P}$) and homogeneity further complicates RF optimization, requiring trade-offs that are anatomy and sequence specific.

The management of B1+ inhomogeneities must be simultaneously addressed with local heating concerns (i.e., peak local specific absorption rate or pSAR). The detailed characterization and validation of RF coils [9] along with innovations like virtual observation points (VOPs) [10] has made it possible to explore pSAR-constrained pTx optimization and the real-time pSAR monitoring necessary to safely and efficiently implement static and dynamic pTx solutions.

Technological innovations

RF coils

While prototype systems with whole-body RF transmitters behind the bore liner have been [11] and are currently in development [12], the majority of 7T systems are configured to use local transmit and receive RF coils. While challenges exist, local transmitters have several advantages, such as increased transmit efficiency and localized electric fields, which simplifies the implant safety approval process.

Local coils for body imaging have been under continuous development in multiple centers worldwide and increasingly by third-party vendors. Coils for proton imaging almost always use multi-channel transmit with 8 to 16 independent channels. The metrics used to evaluate these coils include transmit power efficiency ($B1+/\sqrt{P}$), pSAR efficiency ($B1+/\sqrt{pSAR}$), pTx degrees of freedom, SNR,

parallel imaging performance, and effective coverage. The resonant structures that comprise these arrays consist of loops, micro-striplines, dipoles in multiple configurations, and meander elements, to name a few. Of these, fractionated dipoles [13] and meander elements [14] currently hold the most promise in terms of the depth of penetration and the field profile.

The density of transmit elements (i.e., how many can be packed around a given anatomy) is limited by coil coupling, and it is important with respect to achieving peak transmit B1 fields and degrees of freedom for pTx optimizations. When placing 8–10 dipoles around an average-sized torso at 7T, and 16 dipoles at 10.5T [15], the worst-case coupling is ~10 dB. To further increase the element density, geometrically decoupled loop elements can be paired with each dipole. This strategy is especially beneficial on pTx systems with higher numbers of independent transmit channels (e.g., 16). In our institute, after successful iterations with 8-channel and 16-channel micro-stripline arrays [16], we developed a 16-channel loop-dipole (16LD) transceiver array [17]. While the 16LD performed well on transmit, we were not realizing the expected SNR gains compared to 3T.

While nearly maintaining transmit performance of the 16LD, we targeted increased SNR with a shielded 32-channel loop-dipole array (32LD-SH), which is fully detailed in a recently accepted manuscript [18]. Briefly, the coil is comprised of 8 shielded blocks, each with a fractionated dipole and three overlapping loops. On each block, the dipole is driven with a single RF power amplifier (RFPA), while the three loops are driven by a second RF power amplifier through a three-way splitter. On signal reception, all 32 channels receive independently. The splitter, transmit/receive switches, and preamps are all housed 2 cm above the resonant structures behind an RF shield, which also serves as a barrier to interference from hands or arms being placed on the coil, which can significantly change the S-matrix, thus degrading performance. The coil and the SNR performance versus the 16LD and a typical 3T RF coil configuration are shown in Figure 1. A nearly 3-fold improvement compared to 3T is observed in the periphery of the phantom, and an over 2-fold improvement is seen in the middle. The 32LD-SH was validated for use in multiple anatomies throughout the torso and uses location-specific VOPs for optimal utility while ensuring safe operation.

RF management

The ability to obtain high-quality and reproducible UHF body imaging is directly linked to the successful development and implementation of RF management methods both in terms of field optimization and hardware.

Methods have been classified as static or dynamic pTx [8]. Static methods are characterized by the manipulation

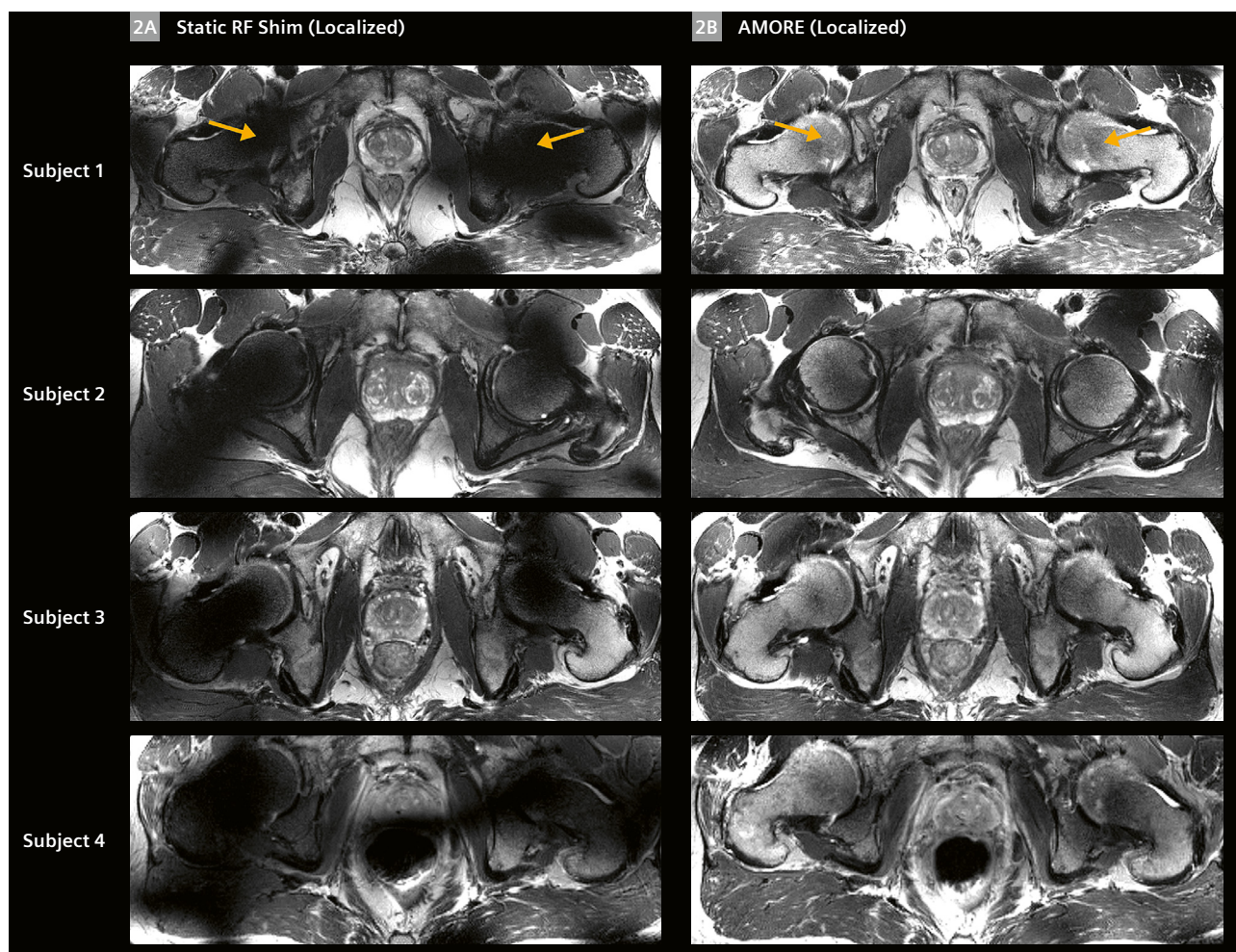
of magnitudes and phases across transmit channels with otherwise identical RF waveforms, while dynamic methods are characterized as having channel-specific RF waveforms. Multiple static and/or dynamic pTx solutions can also be combined by applying distinct solutions to different RF pulses throughout a sequence, as in our previous work on non-contrast renal angiography [7] and perfusion [6], or as in time-interleaved acquisition of modes (TIAMO) [19, 20].

Enabling subject, target, and sequence-specific RF optimization involves acquiring appropriate calibration data. Depending on the pTx strategy employed, different requirements exist for the information content and fidelity of the calibration data. Universally, this data must provide channel-wise B1+ magnitude and/or phase information, with absolute magnitude maps being required for more advanced pTx methods [21, 22]. Relative channel-wise B1+ maps can be obtained quickly (i.e., several slices in

~30 seconds) and converted to absolute maps with a single additional measurement [22, and references therein to Dr. van de Moortele]. However, body applications face challenges due to motion and B1+ inhomogeneity, rendering methods that have been developed for the head less effective. To increase the dynamic range for B1+ mapping, strategies such as B1TIAMO [23] and sandwiched TurboFLASH (TFL) [24] have been proposed.

In vivo applications and performance benchmarks

Achieving optimal RF performance necessitates a tailored approach that considers RF coils, calibration methods, and RF optimization strategies concerning the specific pulse sequence and anatomy. While initial local efficiency static RF shimming targeting the prostate yielded excellent



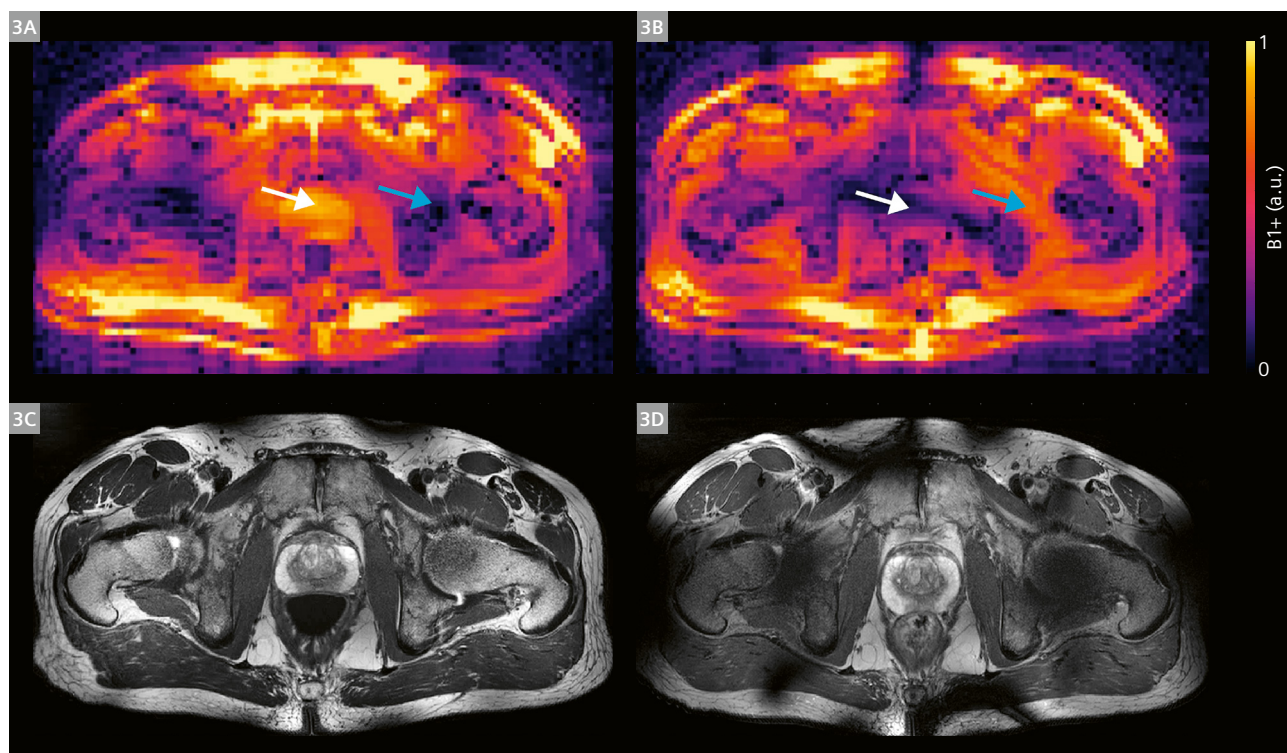
2 Anatomical T2w prostate images in four individuals at 7T with the 16LD-RF coil. **(2A)** Images using a localized static RF shim focusing on the prostate and **(2B)** localized AMORE shims [20]. The AMORE solution consistently provides high-quality T2w imaging of the prostate while also producing higher-quality images over the entire FOV. The nulls in the static RF shim solution are resolved in the AMORE images (e.g., orange arrows in Subject 1). Displayed without image intensity correction.

transmit efficiency and adequate homogeneity for turbo spin echo (TSE) imaging, it resulted in compromised image quality beyond the prostate, with B1+ nulls causing signal voids in the image (Fig. 2A). To improve TSE image quality over a larger target or field of view (FOV), more advanced pTx methods that offer more degrees of freedom are required.

In the TIAMO approach, complementary static RF shims are interleaved with respect to repetition time while repeating the same spatial encoding steps [19]. Assuming two RF shims or modes, the potential 2-fold increase in scan time can be addressed by combining the data acquired from each mode into a set of virtual coils, which can better support doubling the parallel imaging acceleration of a scan. The two modes could be default modes, such as CP and CP2+, or optimized to achieve a uniform root-sum-of-squares (RSOS) of B1+ across the two modes [25]. However, to design modes specifically for TSE imaging, we devised a new optimization strategy referred to as acquisition modes optimized for refocused echoes (AMORE) [20]. This strategy provides more uniform signal and

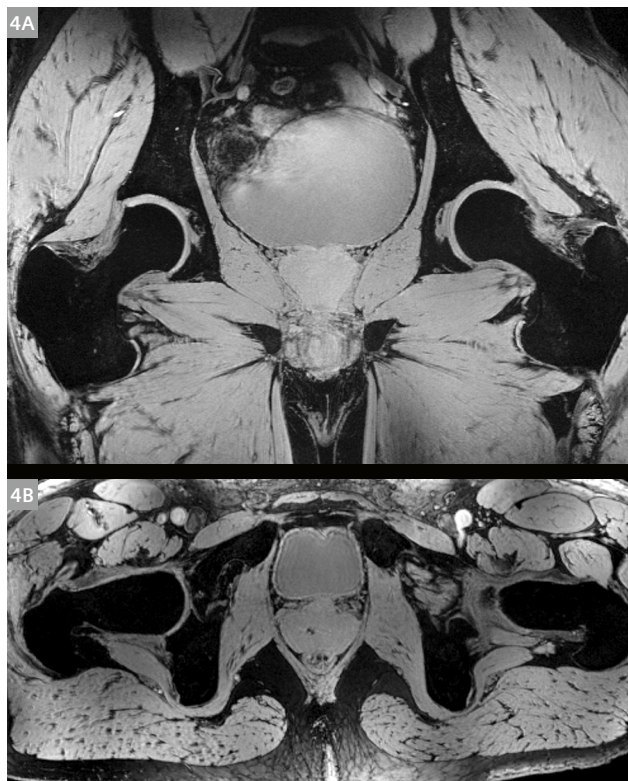
contrast over larger FOVs based on a constrained optimization of the maximum B1+ across the modes. A variation on this method is localized AMORE, where one mode is focused on a target anatomy such as the prostate (Fig. 2B). As seen in the flip angle maps of the two localized AMORE modes, one mode favors the prostate while the other fills in the B1+ nulls in the rest of the torso (Figs. 3A, 3B).

An important finding from these results is the ability to maintain image contrast while halving the repetition time (in the case of a two-mode acquisition) as can be appreciated in the AMORE compared to the static solution shown in figures 3C and 3D. While this reduction in repetition time inherently increases SAR, we leveraged a fully characterized RF coil with VOPs and real-time power monitoring to optimize local SAR hotspots, creating a “SAR hopping” effect and thereby minimizing time-averaged pSAR. This permits localized AMORE to achieve improved full FOV image uniformity and signal within the same scan time as a single static shim, as demonstrated for kidney imaging in abstract #3446 from the 2025 ISMRM Annual Meeting.



3 Prostate imaging comparing a static RF shim with AMORE at 7T using the 16LD-RF coil [18, 20]. Two complementary AMORE modes (i.e., B1+ maps) where the first mode (**3A**) is focused primarily on the prostate (white arrow) and the second mode (**3B**) fills in the B1+ nulls in the rest of the torso (blue arrow). (**3C**) Anatomical TSE imaging of the prostate using the two complementary AMORE modes (3A, 3B) with a TR of 3 seconds. (**3D**) Anatomical TSE image of the prostate using a local efficiency static RF shim localized to the prostate and a repetition time of 6 seconds. While the prostate has high B1+ efficiency with a static shim on the prostate, the B1+ nulls in the larger FOV result in signal dropout and loss of contrast. The signal and contrast in the prostate between the AMORE and static shim images are nearly identical while the AMORE scan maintains signal through the entire FOV in the same scan time.

While originally designed for TSE acquisitions, AMORE also enhances power efficiency for gradient echo (GRE) imaging. Figure 4 showcases this, with two example planes from a 0.67 mm isotropic 3D pelvic acquisition using a two-mode AMORE multiecho GRE sequence with fat suppression. This acquisition strategy was proposed by Scheenen et al. for use with ultrasmall superparamagnetic iron oxide (USPIO) particles to distinguish between healthy lymph nodes and lymph nodes suspicious for harboring metastases [26]. The ability to detect positive lymph nodes



4 Pelvic imaging at 7T with the 32LD-SH coil. **(4A)** Coronal and **(4B)** axial reformats from a high-resolution (0.67 mm isotropic) multi-echo GRE water selective excitation acquired with AMORE [18]. These acquisitions are created by taking the RSOS across echoes after performing the joint reconstruction of the two AMORE modes.

down to 1.5 mm [27] at 7T outperforms molecular imaging methods such as choline positron emission tomography (PET) and ^{68}Ga -labeled prostate-specific membrane antigen (PSMA) PET, neither of which can reliably detect nodal metastases below 5 mm in size [26]. These acquisitions also provide excellent visualization of the articular cartilage for musculoskeletal applications.

Kidney imaging at 7T holds promise for anatomical and functional imaging in chronic kidney disease and cancer, as previously reviewed [3]. Using the 32LD coil, a host of anatomical imaging series were performed to demonstrate the potential of obtaining high-quality kidney images with a strategy as simple as a localized static shim solution (Fig. 5). Vessel imaging in the kidneys was performed using a sequential 2D fat-suppressed GRE sequence. The longer T1 of tissue at higher field strength results in more pronounced background tissue suppression for a given sequence timing, and thus more in-flow enhancement as previously exploited for time-of-flight angiography in the head. Anatomical imaging of the kidneys was acquired with both TSE and fat-suppressed multi-slice GRE acquisitions.

Due to the larger volume of the liver, more degrees of freedom are needed to obtain a uniform signal and contrast across the whole organ. The data shown in Figure 6 includes anatomical liver imaging at 7T acquired with 2D multi-slice GRE using two-mode AMORE. At 10.5T, we demonstrated the use of multi-spoke RF pulses to acquire similar data, resulting in relatively uniform contrast across the liver [28] (Fig. 7). By manipulating the excitation k -space, multi-spoke RF pulses are optimized to provide a more uniform excitation, whereas TIAMO methods use complementary fields to generate a spatially uniform signal or contrast.

For cardiac imaging, we and others have previously shown the ability to measure quantitative functional parameters at 7T that were equivalent to 1.5T and 3T [29, 30]. An example of the image quality obtainable with the new 32LD-SH coil for functional cardiac CINE imaging used for these assessments is shown in Figure 8. Beyond

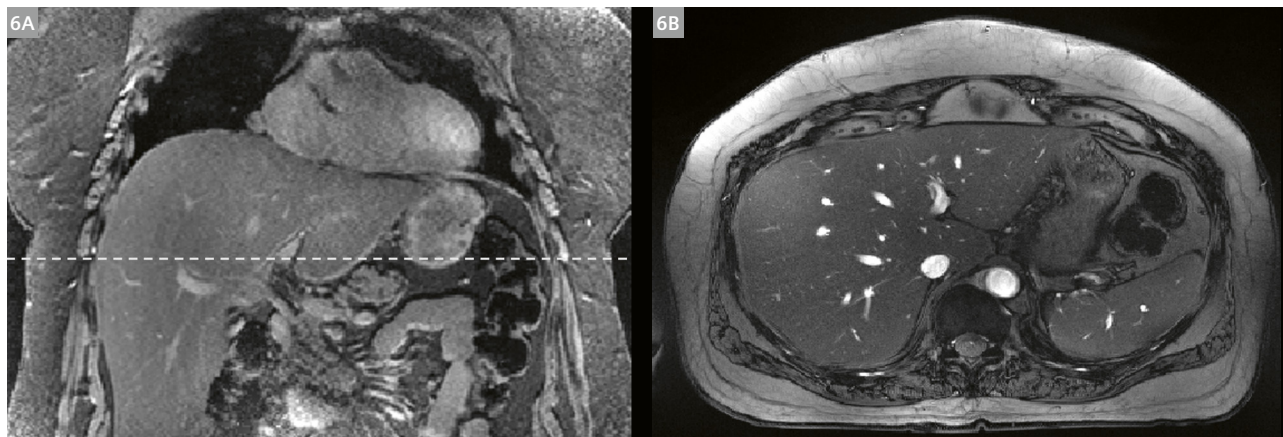


5 Renal imaging at 7T with the 32LD-SH coil acquired using fat suppression and a static efficiency shim over both kidneys and renal arteries, including: **(5A)** T1w GRE anatomical imaging, **(5B)** multi-slice in-flow imaging shown as a 20 mm maximum intensity projection, **(5C)** TSE imaging showing renal anatomy highlighting the high-intensity signal from the urine-filled renal calyces [18].

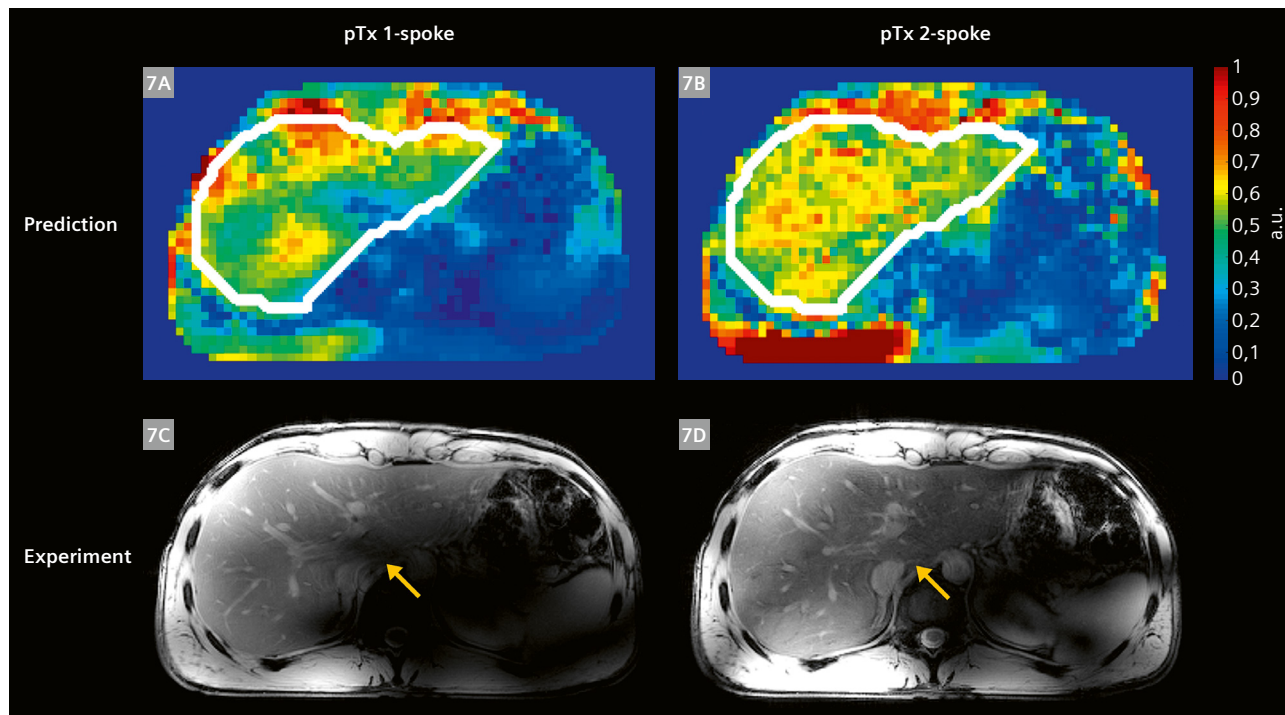
this, the potential improvement in spatial and temporal resolution promises improved evaluation of function, flow, structure, and metabolism in the diagnosis and monitoring of cardiomyopathies and coronary artery disease. Current studies demonstrating more routine use of 7T for cardiac examinations highlight that the technology is close to being realized in the clinical setting [31].

Outlook for clinical adoption

There is tremendous momentum to realize body imaging in the clinical setting at UHF¹ as more 7T systems are installed in university hospitals and destination imaging centers. Despite the many successes, hurdles to clinical adoption remain and can be broken down into four main



6 Anatomical images of the liver at 7T with the 32LD-SH coil. A coronal GRE scout (**6A**) is shown with a dashed line indicating the position of the axial image in (**6B**). The axial image is a receive field corrected anatomical T1w image at the level of the liver and spleen using TIAMO for RF management.



7 Liver imaging at 10.5T showing the advantage of the pTx solution with increased degrees of freedom and the field strength and target anatomies getting larger [28]. The predicted B1+ profiles (**7A**, **7B**) and acquired low-flip-angle GRE images (**7C**, **7D**) in liver with different RF management strategies: (**7A**, **7C**) 1-spoke pTx pulse; (**7B**, **7D**) 2-spoke pTx pulse. The acquired images matched well with predictions in terms of the images generated by the predicted field maps. The increased degrees of freedom of the 2-spoke pTx pulse effectively mitigated the B1+ inhomogeneity in the liver as demonstrated by improved consistency in contrast throughout the organ (orange arrows). No receive field correction was applied.



8 Cardiac imaging at 7T with the 32LD-SH transceiver array. Single frames from cardiac CINE acquisitions in the **(8A)** short-axis, **(8B)** 4-chamber, and **(8C)** 2-chamber orientations all obtained with a CV regularized efficiency shim. Standard cardiac views were acquired with retrospective gating and reconstructed with 50 phases.

categories: technical, implementation, regulatory, and acceptance.

Many of the **technical** hurdles have been delineated above and are being addressed in the research community. First is the configuration of the transmit chain available on the MRI system. With the assumption of full pTx functionality, 8-channel transmit systems are typically available, with a few 16-channel transmit systems in the field. The transmit architecture must work seamlessly with the RF coil to deliver the performance needed to run all necessary imaging sequences for clinical diagnosis. Finding the ideal coil to realize the promised SNR gains while maintaining the needed transmit performance is also key with local transceiver array coils. Pursuing the alternative strategy of a behind-the-bore transmitter is also an intriguing option as it mirrors the functionality of 1.5T and 3T closed bore systems. The cost-benefit tradeoff needs to be investigated, considering the increased expense of the RF coil and transmit chain, and balancing it with the decreased cost and potential benefits of local receive-only array technology.

With respect to **implementation**, there are many issues related to ensuring robust and integrated functionality to enable the use of advanced features in a clinic-friendly workflow. First, we know that this will involve pulse sequence modifications to incorporate the needed pTx functionality, whether static, dynamic, or some combination of the two. It may also involve modifications of RF pulses and management strategies to handle limitations in peak B1+, especially when considering the known variability in patient anatomy. Coupled with this would be a suite of sequence- and target-specific calibration and RF management solutions to streamline pulse sequence optimization and utilization. One such strategy would be to develop universal pulses and static shims to use for studies or as a starting point for subject-specific calibrations. For example, we have shown that developing calibration-free universal shims and TIAMO modes is advantageous in the heart and pelvis at 7T and 10.5T, respectively [32]. This

work identified tailoring universal solutions to sub-populations based on BMI, which further improved performance simply by knowing a subject's height and weight.

On the **regulatory** front, it is clear that the approval of body imaging applications will depend on sufficiently addressing the above technical and implementation hurdles prior to running patient studies. If possible, the goal of showing equivalence to lower field strength systems should be the initial goal, with the knowledge that its clinical benefit will be learned through patient imaging and the process of active discovery. This is, of course, assuming that robust, high-quality data can be routinely acquired without requiring a team of physicists. Meeting these goals will improve overall **acceptance** by setting realistic expectations and allowing clinical users to explore how to best use the advantages of 7T to improve patient care. Perhaps this will come in the form of higher resolution scans for prostate and hip imaging, more sensitive non-contrast flow or perfusion studies in the heart or kidney, or increased sensitivity to oxygenation and exogenous contrast agents for diagnosing and staging cancer. While proton imaging provides the foundation for most clinical work, the increased SNR can be leveraged to boost the sensitivity to low-gamma nuclei as well (^{23}Na , ^{31}P , ^2H , ^{13}C , etc.). The possibility of providing highly sensitive and comprehensive anatomical, functional, and metabolic exams should motivate the clearing of hurdles preventing the realization of torso imaging in the clinic at 7T.

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Whole-Body Evaluation of the Bone using PD-Weighted VIBE Sequences with Deep Learning Reconstruction and Applications in Oncology

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In the last decade, whole-body MRI (WB-MRI) has been established as a standard diagnostic and treatment-monitoring tool for cancer patients with bone disease [1, 2]. Bone metastases are a common finding in late-stage cancer patients and can be associated with poorer prognosis, treatment resistance, and tumor-related adverse skeletal events, such as fractures, spinal cord compression, and the need for bone radiotherapy.

Standard WB-MRI protocols, as prescribed in the Metastasis Reporting and Data System (MET-RADS) and the Myeloma Response Assessment and Diagnosis System (MY-RADS) [3, 4], focus exclusively on the evaluation of bone marrow involvement, largely because of the inability to visualize the mineralized bone. Thus, WB-MRI has always had the limitation of requiring computed tomography (CT) scans to confidently evaluate the bone structure for surgical interventions and radiation therapy planning.

In recent years, there has been increasing interest in the development of CT-like images using MRI. Acquiring CT-like images simultaneously with conventional WB-MRI could have several benefits, including negating additional

CT appointments, eliminating ionizing radiation doses, and minimizing spatial registration errors. Indeed, the creation of synthetic CT-like images for radiotherapy planning purposes is already possible using the *syngo.via* RT Image Suite. However, these images cannot provide the level of bone matrix detail needed for the radiological diagnosis of bone pathology.

Multiple investigators have reported bone MRI using a variety of VIBE and ZTE/UTE approaches [5, 6]. Structures with no signal return show black against a uniformly gray soft tissue background—known as “black-bone” imaging. When the image contrast is inverted, the bone appears white, like that in CT.

Recently, we reported that Deep Resolve Boost (DRB) reconstruction could cut whole-body diffusion-weighted imaging (WB-DWI) sequences by almost 50% without corresponding declines in image quality [7]. Building on our experience, we developed and clinically implemented a proton density (PD)-weighted VIBE sequence¹ using DRB to demonstrate its added value for bone imaging in WB-MRI examinations.

¹Work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

Black-bone contrast with VIBE

We iteratively developed 2 mm slice thickness StarVIBE and Cartesian VIBE sequences with a PD-weighted, 3° black-bone flip angle. A low flip angle was chosen to reduce the soft tissue contrast as much as possible, mimicking bone-window CT, while maintaining a workable SNR.

Images were acquired in-phase to reduce the appearance of the chemical shift artifact of non-bone structures. Receive bandwidth can be increased to reduce pixel shifts further as desired. Regardless of parameter selection, air and other non-signal-generating anatomical structures (e.g., bowel air, metal implants, surgical clips) appear as high signal on inverted images and can be distracting to radiologists.

StarVIBE is robust against motion artifacts due to the radial *k*-space sampling method, and produces good-quality CT-like images. This technique relies on setting an optimal number of radial views that are sufficient to fill the *k*-space for the selected matrix. The acquisition time (TA) increases with the number of radial views and slices, which limits the practicality of applying this method across the whole body.

SNR was noted to be extremely poor with conventional VIBE sequences, owing to the selected flip angle. We applied Deep Resolve Sharp and Boost-enhanced VIBE (DL-VIBE)¹, provided by Siemens Healthineers as a research application, to improve image quality. DL-VIBE can be acquired either with breath-hold for imaging of the mobile skeleton (e.g., ribs, sternum) or under free breathing for higher resolution. We used DL-VIBE as the preferred approach because it provides adequate bone details for diagnosis within an acceptable total acquisition time. The key parameters for StarVIBE and DL-VIBE are shown in Table 1.

Whole-body applications

Vertex-to-thigh body coverage can be achieved in less than 5 minutes using breath-hold DL-VIBE. Paired with the reported time savings of DRB WB-DWI [7], it may be possible to integrate this sequence within 60-minute appointment times.

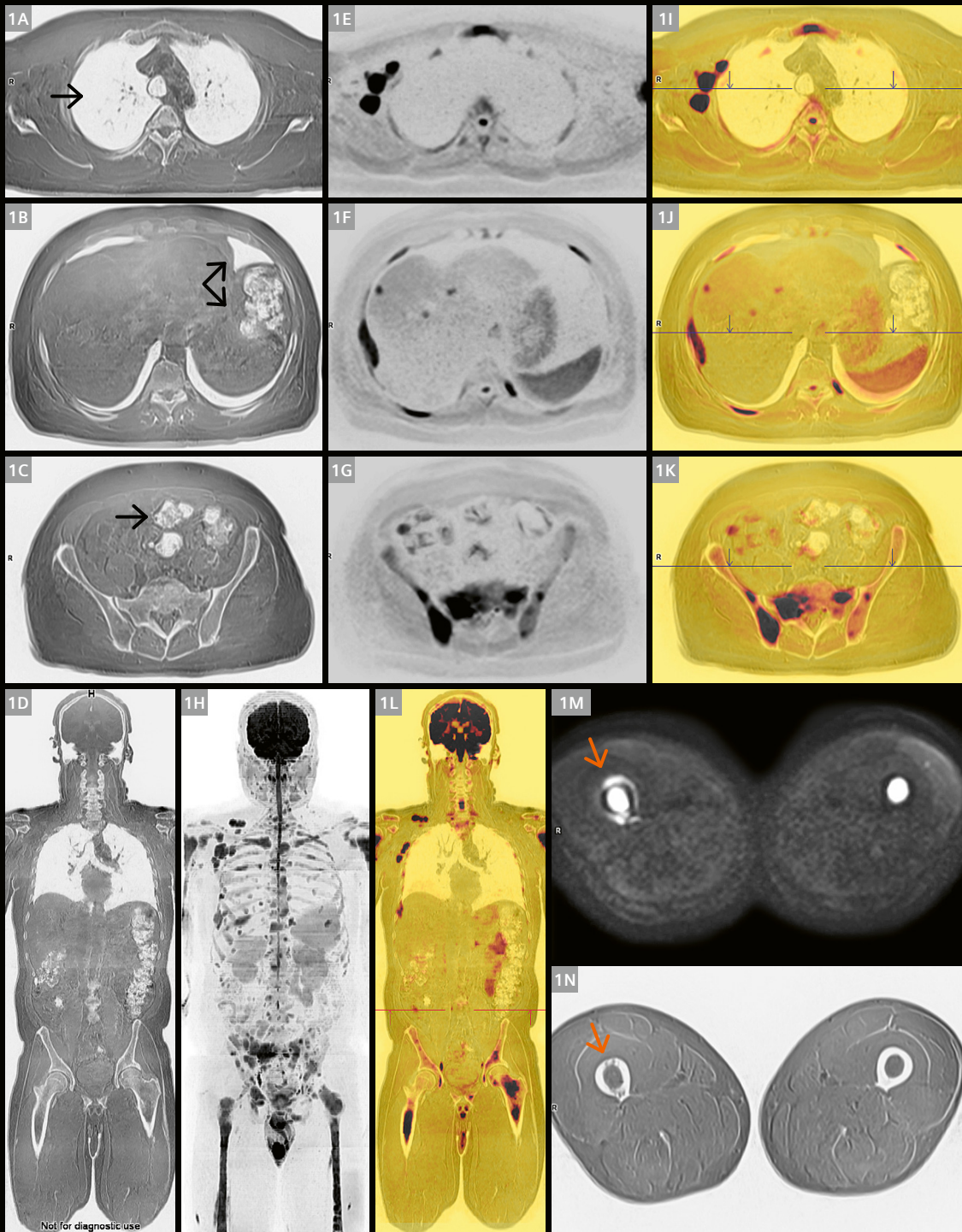
Composing stack-of-slice groups can result in stepping artifacts along continuous bone structures, unless 3D distortion correction is applied. This correction results in signal voids in several of the first and last slices of a group. Therefore, it is important to ensure that sufficient overlap is applied. Nevertheless, we have noted that a 2 mm slice thickness facilitates good quality multi-planar reformatting (MPR), which is useful for whole-body and long-bone evaluations as illustrated in the cases.

The CT-like images can be fused with $b = 900 \text{ s/mm}^2$ DWI sequences to produce images with a similar appearance to PET/CT. Fusion imaging increases the conspicuity of hypercellular deposits and clearly communicates the positions of abnormalities in the skeleton. From PET/CT and SPECT/CT, we know that the CT component adds to the specificity of diagnoses by reducing false-positive findings [8]. We are investigating if the CT-like VIBE MRI provides similar benefits for WB-MRI examinations.

Sequence	Slices	TE (ms)	TR (ms)	Flip angle (degrees)	Field-of-view (mm) / matrix	k-space trajectory	Acceleration factor	Partial Fourier	Acquisition time per slab (min:sec)	Technical notes
StarVIBE	176 @ 2 mm	4.77	6.61	3	400 / 256	Stack-of-stars; 320 radial views	N/A	Slice 5/8	06:00	Stellate artifact from the radial trajectory
Breath-hold DL-VIBE	176 @ 2 mm	4.77	6.4	3	400 / 224	Cartesian	3 (PE) 2 (3D)	Phase 6/8 Slice 6/8	00:17	Good image quality, sharpness limited by breath-hold capacity
Free-breathing DL-VIBE	176 @ 2 mm	4.77	6.54	3	400 / 384	Cartesian	2 (PE) 2 (3D)	Disabled	02:04 (2 averages) 01:03 (1 average)	Resolution is comparable to CT. 2 averages optimal for elevated BMI

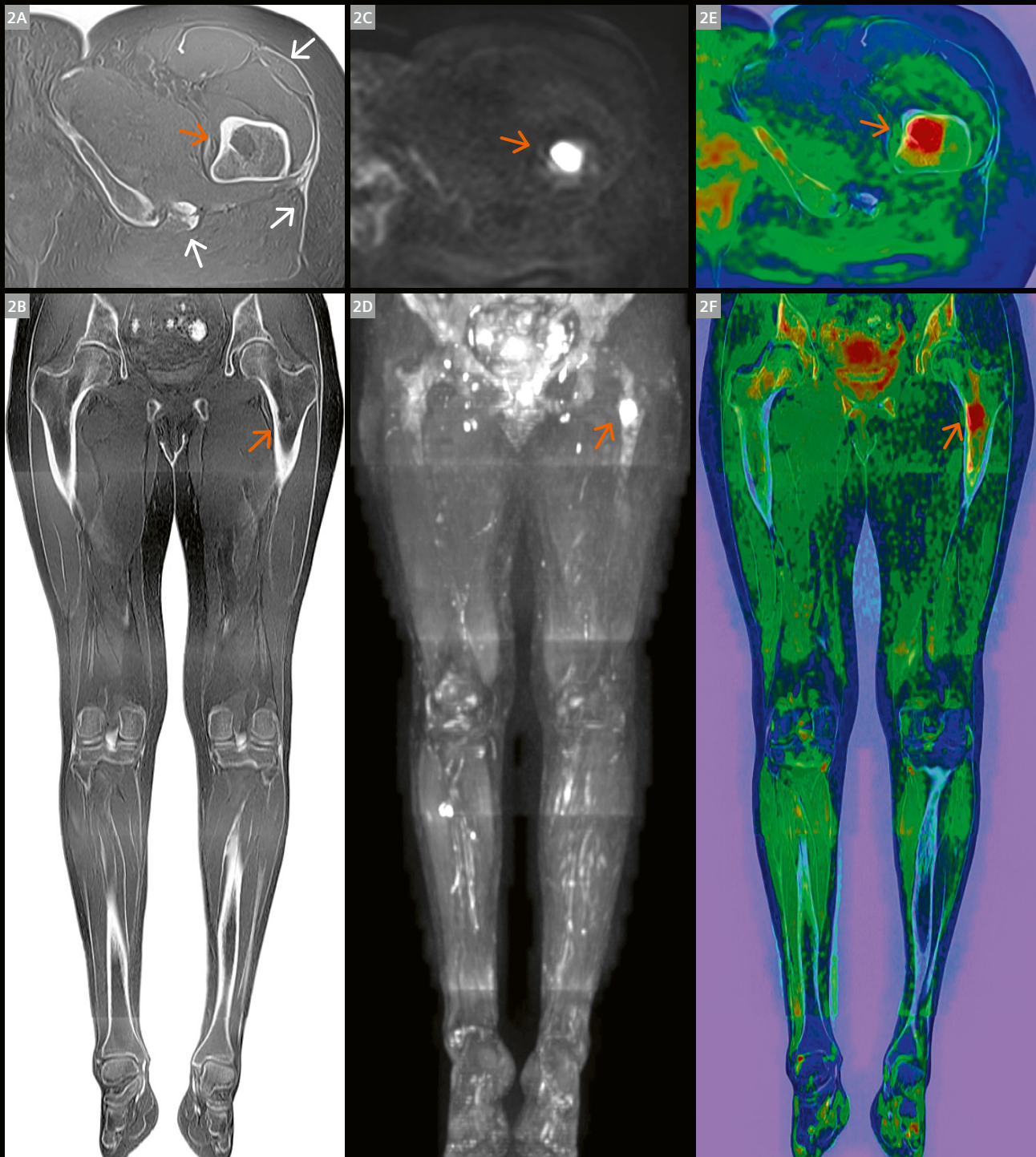
Table 1: Key parameters for StarVIBE and DL-VIBE on a 1.5T MAGNETOM Sola.

Case 1



- 1** A 54-year-old male patient with multiple myeloma post autologous transplant. Multiple hypercellular bone lesions are seen throughout the skeleton and in the right axillary lymph nodes. Breath-hold DL-VIBE (1A–1C, 1D) and inverted b900 (1E–1G, 1H) were acquired across a whole-body range in the axial plane. Fused DL-VIBE/inverted b900 (1I–1K, 1L) enable precise registration of bone marrow disease in relation to the cortical bone. Coronal MPR of DL-VIBE (1D) demonstrates good-quality composed images with no artifactual stepping of continuous bone structures. Note the artifactual brightness of non-signal-returning lung and bowel gas (black arrows). Extra-osseous, paramedullary disease (orange arrows) was observed on b900 (1M) with DL-VIBE (1N) demonstrating an intact bony cortex.

Case 2



2 A 59-year-old female patient with metastatic melanoma. Higher resolution free-breathing DL-VIBE (2A), b900 (2C), and fused DL-VIBE/b900 (2E) using an alternative color look-up table. Coronal MPRs and non-inverted MIP projection are shown (2B, 2D, 2F). Note the artifactual bright low-signal-returning tendons and muscle fascia (white arrows). A pericellular lytic lesion is seen in the left femur (orange arrows). With DL-VIBE, it was possible to confidently determine that the overlying cortical bone is intact.

Conclusion

Deep learning reconstructions, specifically Deep Resolve Boost, can markedly shorten the acquisition times of PD-weighted VIBE sequences to produce CT-like images. It is possible to undertake this on a whole-body basis and we have successfully integrated it into our routine WB-MRI protocols that comply with MET-RADS and MY-RADS.

Visit

magnetomworld.siemens-healthineers.com/hot-topics/quantitative-whole-body-mri to download black-bone protocols for 1.5T.



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Pregnancy and MRI of the Appendix

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Abstract

Abdominal pain in pregnant patients poses a diagnostic challenge. Acute appendicitis can occur during pregnancy, as it can in any other patient. In contrast to other patients, however, adequate diagnosis and treatment is important to prevent possible complications such as preterm labor. Physical examination and standardized diagnostic procedures (e.g., abdominal ultrasound) are more difficult because of the growing uterus. Due to the clinical consequences, there is a need for further examination in cases of inconclusive diagnostics. In pregnant patients, MRI may provide a solution. This is a highly reliable tool to primarily unveil signs of acute appendicitis and subsequently determine the need for further treatment. In MRI, various sequences including diffusion-weighted imaging (DWI) and derived apparent diffusion coefficient (ADC) maps are helpful to diagnose acute appendicitis and avoid the need for intravenous contrast.

Introduction

Acute abdominal pain is one of the most common reasons for a visit to the emergency department (ED). Acute appendicitis (AA) is a frequently seen diagnosis in patients with acute abdominal pain, most often in the right lower quadrant (RLQ). In 2021, the global age-standardized incidence rate was estimated at 214:100,000 [1]. In ED visits, ultrasound (US) is the most commonly used modality to gain information about the appendix and its condition, due to the lack of ionizing radiation and its widespread availability. However, cases in which US cannot rule out AA are not uncommon. They require additional imaging studies, most often a computed tomography (CT) exam.

In pregnant patients with acute abdominal pain, approximately 66.7 to 80:100,000 have AA [2]. The physical examination of the abdomen is more difficult during pregnancy because of altered organ placement brought about by the gravid uterus [3, 4]. The appendix tends to migrate superiorly, resulting in different findings in the clinical assessment and less distinct presentation of the classic RLQ pain [5]. Laboratory indicators used for diagnosis are less reliable during pregnancy. Additionally, the US exam is complex because the growing uterus limits visualization

This article is written in the concise Compendium Medicine style, using *bells* to outline important warning signs and clinical symptoms, *light bulbs* for interesting facts, and a *condition summary* using the first letter of each indicated step in the clinical process/patient follow-up (see the abbreviation list).

Abbreviations

AA	Acute appendicitis
ADC	Apparent diffusion coefficient
CT	Computed tomography
D	Definition
DDx	Differential diagnosis
DWI	Diffusion-weighted imaging
Dx	Diagnostics
E	Epidemiology
ED	Emergency department
Et	Etiology
GBCAs	Gadolinium-based contrast agents
Hx	Patient history
IV	Intravenous
MRI	Magnetic resonance imaging
NOM	Non-operative management
P	Prognosis
PE	Physical examination
RLQ	Right lower quadrant
RUQ	Right upper quadrant
Tx	Treatment
US	Ultrasound
!	Watch out/don't forget

of the appendix during pregnancy. Imaging using CT is not recommended due to the fetal risks, which include slow growth and development, malformations, and impaired brain function, depending on the radiation dose [6]. In pregnant patients with suspected AA, magnetic resonance imaging (MRI) may be necessary to confirm the diagnosis and help with clinical decision-making in cases where repeated US is inconclusive.

Clinical dilemma

During pregnancy, the diagnostic trajectory of acute abdominal pain is more difficult due to the range of possible causes and the increased difficulty of assessing anatomical structures. Additionally, diagnostic imaging may be accompanied by fetal risks, but AA is also accompanied by fetal risks (e.g., premature labor, intrauterine fetal demise) [7]. In case of inconclusive US in a pregnant patient, MRI is the modality of choice to identify AA and define treatment options. But is it safe? And is intravenous (IV) contrast needed?

Diagnostic imaging

Conventional imaging in a pregnant patient can pose a challenge. Imaging of appendicitis, especially at an early stage can be difficult, even in a slim nonpregnant patient. When inflammatory markers are low, secondary signs of inflammation cannot always yet be appreciated, regardless of the chosen modality. When an US exam is performed, secondary signs such as fat stranding, free fluid, and possibly (in cases of visualization of the appendix) wall thickening and an appendicolith may guide a radiologist to the diagnosis of AA [8]. However, in otherwise “healthy” individuals, the appendix may be hidden behind the cecum and difficult to assess due to intraluminal air. In pregnant patients, this is additionally challenging due to displacement of the cecal pole and appendix out of the iliac fossa and subsequent posterior tilting.

In patients revealing increasing inflammatory markers, persistent abdominal pain, and subsequent persistent suspicion of AA, treatment may be preferred. However, especially in pregnant patients, invasive treatment must be avoided unless deemed necessary. Depending on clinical signs, reevaluating the next day may be considered as causing an unnecessary delay in pregnant patients. In these cases, prompt diagnosis is required, and an MRI exam may be the tool of choice.

An MRI exam for appendicitis can include a wide range of imaging sequences, which is reflected in the many different protocols that vary between clinics [8]. Sequences that may be specifically beneficial are T2-weighted and T2w fat-saturated images to visualize the periappendicular edema and free fluid. Preferably, these images are reconstructed in three planes to enable finding the appendix. Additional sequences that may be specifically beneficial are diffusion-weighted imaging (DWI) and derived apparent diffusion coefficient (ADC) maps.

The DWI technique enables visualization of the Brownian motion of water molecules within a voxel of tissue. Tissues with a high cellular count or cellular swelling

(i.e., inflamed tissue) will allow less Brownian motion. This results in lower diffusion coefficients (i.e., diffusion restriction). To avoid misinterpretation, DWI always needs to be interpreted together with the ADC values. A high signal on the DWI may be due to diffusion restriction or due to the T2 shine-through effect, where high T2 signal intensity results in an artifactual high signal on the isotropic maps. Actual diffusion restriction will result in high signal intensity on the DWI and low signal intensity on the ADC map [9].

As explained above, DWI will reveal restricted diffusion in the appendix when inflammation is present. If the appendix cannot be found on the T2 reconstructions, the DWI may function as your roadmap. Periappendicular free fluid and restricted diffusion combined with clinical findings are sufficient to diagnose AA.

As diffusion imaging can provide an answer to the clinical question, IV contrast in a pregnant patient with suspected AA is not required. In cases of suspected complicated AA (i.e., with a periappendicular abscess), the need for IV contrast in a pregnant patient may be discussed, as the DWI may indicate the abscess as well. Therefore, IV contrast should only be considered if it is likely to lead to a different conclusion with different clinical consequences (regarding the choice of therapy).

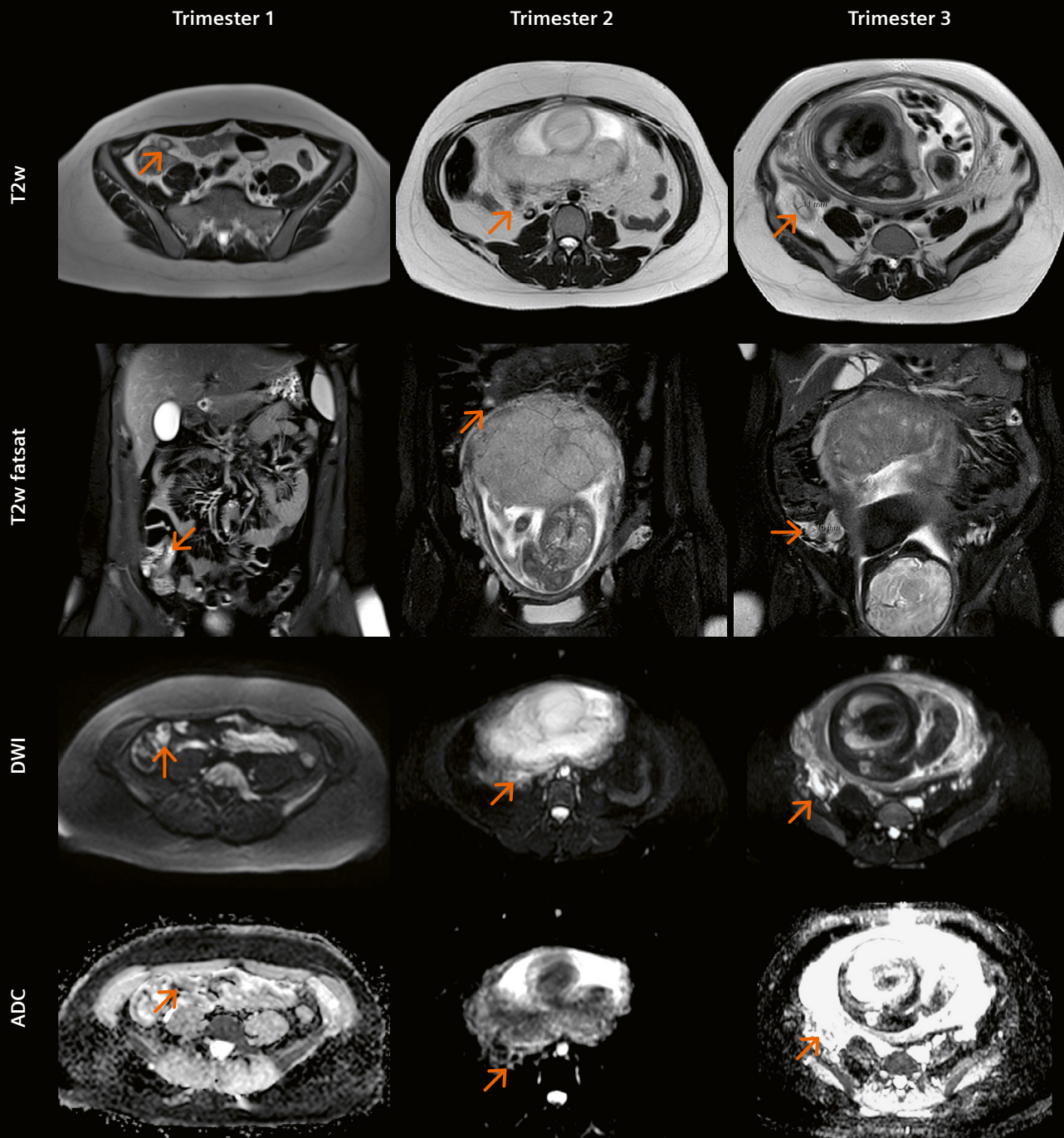


Gadolinium-based contrast agents are known to cross the placental barrier [10]. They are found in the fetal circulation and in the amniotic fluid after renal excretion. They are also known to leave deposits in the brain [11, 12].

Gadolinium-based contrast agents

Gadolinium-based contrast agents (GBCAs) are the most commonly used contrast agents in MRI exams. Although MRI itself is considered safe after 18 weeks of pregnancy, it is not proven safe and should therefore only be performed in cases of clinical consequences. Possible risks for the fetus are not fully known when imaging is performed during the first trimester, but the probable risks reduce with the strength of the magnetic field, i.e., the risk is lower with lower field strengths scanners [10].

Safety concerns about the use of GBCAs in pregnant patients have been raised. With current, limited data, the safety of GBCAs remains unclear. The 2021 guideline from the American College of Obstetricians and Gynecologists suggests using GBCAs if they will significantly increase diagnostic performance and are expected to improve maternal or fetal outcomes, outweighing possible risks [13, 14]. Scientific research on this topic is difficult due to ethical considerations, making it unlikely that evidence-based recommendations will follow in the foreseeable future.



- DWI: sensitivity 99%
- ADC: sensitivity 97%–99%, specificity 97%–99% [15–17]

Acute appendicitis		Acute appendicitis during pregnancy
D Definition	Acute appendicitis is an acute inflammation of the vermiform appendix.	
E Epidemiology	Incidence: 229.9:100,000 per year (worldwide), mostly between the ages of 5 and 45 [22].	• Prevalence similar to that in nonpregnant patients. • Occurs in 66.6 to 80:100,000 pregnancies [2].
Et Etiology	Obstruction of the appendiceal lumen → stasis of mucus → distention of serosa → vascular congestion → ischemia → necrosis → perforation → peritonitis	
Hx Patient history	Initial diffuse periumbilical pain (visceral peritoneum irritation) localizing to the right lower quadrant (RLQ, parietal peritoneum irritation) within ± 12–24 hours, anorexia (80%), nausea/vomiting, fever (40%), diarrhea, malaise, urinary frequency or urgency.	
		May present with milder, subtler symptoms, especially late in pregnancy, localizing to RLQ, right flank, or even RUQ.
PE Physical examination	• McBurney's point tenderness: RLQ guarding. • Rovsing's sign: RLQ pain elicited by palpation of the left lower quadrant. • Psoas sign: pain elicited on flexing the right hip with stretched leg against resistance. • Obturator sign: RLQ pain on passive internal rotation of the right hip with the hip and knee flexed.	• Alders sign [23–25]: area of maximum tenderness is the same when the patient lies on her back versus on her left side. When she is turning, keep applying pressure. • Bryan sign [24–26]: pressure from the left side on the pregnant uterus elicits pain in RLQ or middle quadrant. For both signs, the sensitivity and specificity remain unclear.
DDx Differential diagnosis	• Gastrointestinal: gastroenteritis, cholecystitis/cholecystolithiasis, pancreatitis, diverticulitis, gastrointestinal perforation, ileus, irritable bowel syndrome, bowel ischemia. • Urinary tract: urinary tract infection, pyelonephritis, urolithiasis. • Gynecological: menstrual-/ovulation pain, ectopic pregnancy, pelvic inflammatory disease, miscarriage.	
	N/A	• Placental abruption • Uterine rupture • Pre-eclampsia • HELLP syndrome • Round ligament syndrome
Dx Diagnostics	• Traditionally a clinical diagnosis. • Labs: leucocytes ↑, CRP ↑, urinalysis (usually normal), urine/serum B-hCG (usually normal). • Abdominal ultrasound: distended appendix (> 6 mm diameter), possible appendicolith, and abnormally increased echogenicity of the peri-appendiceal fat. • CT scan preferably with IV contrast: enlarged appendix (> 6 mm diameter), appendiceal wall thickening (> 2 mm), peri-appendiceal fat stranding, appendiceal wall enhancement, possible appendicolith. • Consider MRI in young infants and pregnant patients, in sick patients or for persistent suspicion whilst repeated ultrasound remains inconclusive.	• Clinical diagnosis more difficult due to displacement during pregnancy. • Labs: leucocytes ↑ (can be due to pregnancy), CRP ↑ (can be due to pregnancy), urinalysis (usually normal), urine/serum B-hCG (elevated). • Abdominal US: similar to non-pregnant patients. • MRI if US is inconclusive: periappendicular free fluid, diffusion restriction.
Tx Treatment	• Nil per os. • IV fluids, AB IV, analgesics. • Appendectomy (gold standard), laparoscopy preferred over open approach [27].	
P Prognosis	Mortality rate in developed countries (0.09%–0.24%) varies in resource-limited countries (1%–4%).	
! Watch out / don't forget	Risk of perforation → peritonitis → sepsis	Preterm labor, birth or fetal loss [28, 29]

Treatment

The gold standard in the treatment of AA is an appendectomy. Over the years, laparoscopic appendectomy has become the preferred method due to its minimally invasive nature, shorter hospital stay, and lower complication rates. An open appendectomy is equally effective, yet less favored due to longer recovery times and higher complication rates [18, 19]. There is a growing interest in non-operative management (NOM), such as antibiotics as a primary treatment for uncomplicated appendicitis. However, due to significantly lower treatment efficacy, an appendectomy is still the gold standard [20].

The treatment in pregnant patients is similar to that in non-pregnant patients, with additional considerations for maternal and fetal safety [21]. In pregnant patients, the NOM is associated with higher rates of preterm labor and delivery, and is therefore not recommended. In the treatment of AA, there should be no delay, since each day of delay increases the risk of preterm labor [28, 29].

Conclusion

In summary, the diagnosis and management of AA during pregnancy presents unique challenges due to the anatomical changes and the need to minimize fetal risks associated with imaging. MRI has proven to be a valuable diagnostic tool, especially when US results are inconclusive, offering high sensitivity without the use of harmful ionizing radiation. Both DWI and ADC sequences provide particularly useful insights into the presence of inflammation. Although MRI is generally considered safe after the first trimester, the safe use of GBCAs remains unclear. Therefore, the use of contrast should only be considered when it significantly enhances diagnostic performance and has a substantial impact on clinical decision-making and maternal or fetal outcomes. Prompt diagnosis and careful consideration of treatment options are critical to optimizing outcomes for both mother and fetus.

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For MAGNETOM systems, the following disclaimer is valid: To date, there is no scientific proof that MR examinations are harmless for pregnant women, the unborn (embryos or fetuses), and children under two years of age. An embryo or a fetus is especially sensitive to potential thermal events during the first three months of pregnancy. Therefore, avoid scanning pregnant patients especially in the first three months of pregnancy and avoid scanning patients with unknown pregnancy status.

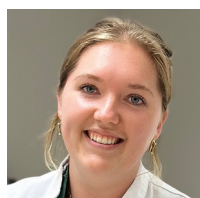
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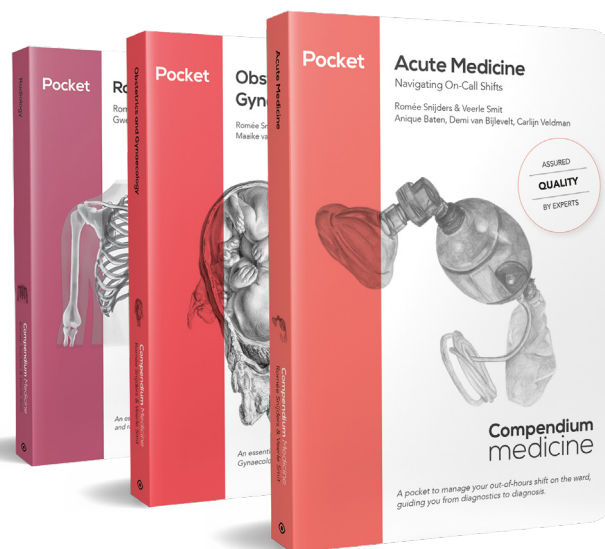
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Sedation-Free Pediatric Neuroimaging with Deep Learning-Reconstructed Deep Resolve 3D MR Sequences

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Introduction

In this article, we evaluate whether 3D sequences with deep learning (DL)-based image reconstruction can replace 3D sequences without DL for pediatric¹ neuroimaging, and help achieve sedation-free MRI exams.

At Oslo University Hospital, we encounter a significant number of pediatric cases, which present unique challenges due to their specific needs. Sedation or general anesthesia are commonly used during MRI exams for younger patients to ensure they remain still throughout the scan.

Cutting-edge AI-based techniques like DL-based image reconstruction have significantly contributed to faster scans, which are crucial for every patient group [1–4]. Faster scans offer the advantage of producing clearer and more detailed images by reducing the occurrence of motion artifacts, leading to more accurate diagnoses and higher success rates of awake MRI scans [5]. Deep Resolve is a newly launched deep learning application that enables faster, more motion-robust scans. Deep Resolve Boost is a powerful denoising algorithm to mitigate noise enhancement due to higher acceleration factors. Deep Resolve Sharp is a DL-based interpolation to improve the image resolution by a factor of two along each spatial dimension.

When scanning younger children¹, this can help reduce sedation rates, along with the associated risks and costs [6]. It also minimizes patient discomfort and can potentially improve the overall patient experience.

Findings and procedure details

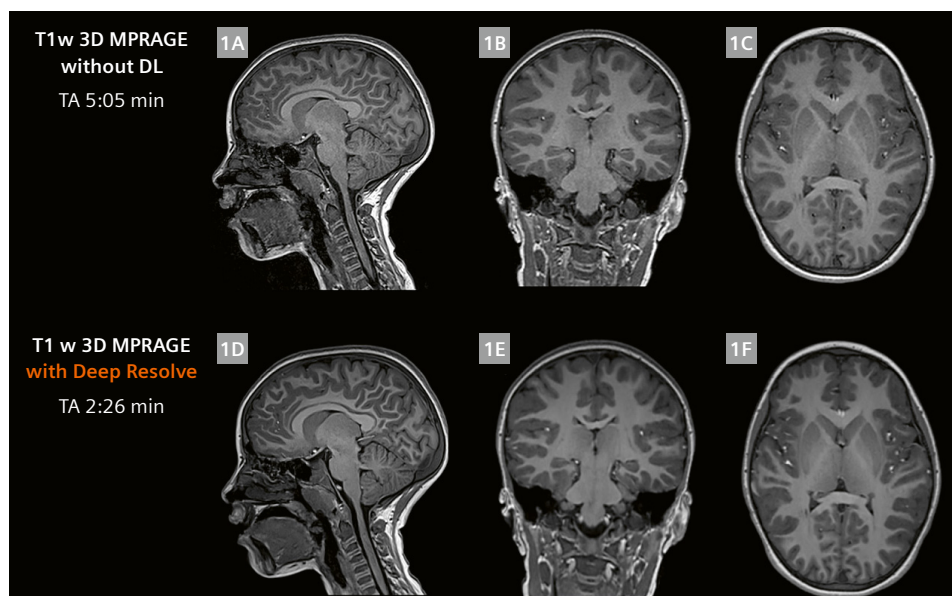
In addition to fast, motion-robust protocols such as Deep Resolve, proper patient preparation is a key element in enabling pediatric MRI without sedation or general anesthesia. Most of our neuro protocols include both 2D and 3D sequences. At our 3T MAGNETOM Skyra Fit (Siemens Healthineers, Erlangen, Germany), Deep Resolve is currently available for 2D sequences, and has been highly beneficial [7]. However, we are eager to have deep learning capabilities for 3D sequences. Currently, we are collaborating with Siemens Healthineers to implement work-in-progress (WIP) sequences² to use Deep Resolve in our 3D sequences. The two most commonly used and time-consuming 3D sequences are T1w 3D MPRAGE and T2w 3D SPACE FLAIR. Applying DL to these sequences would greatly assist in situations where patients struggle to remain in the scanner for extended periods.

¹MR scanning has not been established as safe for imaging fetuses and infants less than two years of age. The responsible physician must evaluate the benefits of the MR examination compared to those of other imaging procedures.

²Work in progress. The applications are currently under development and are not for sale in the U.S. and in other countries. Their future availability cannot be ensured

Case 1

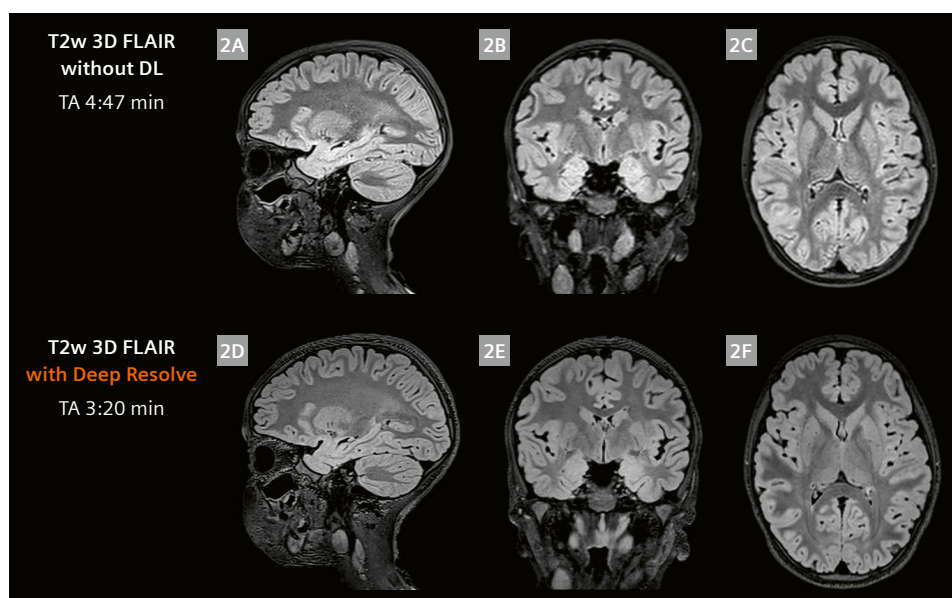
A 6-year-old girl with sickle cell anemia underwent an MRI of the brain prior to stem cell transplantation.



1 T1w 3D MPRAGE

This figure illustrates the difference between T1w 3D MPRAGE images with and without Deep Resolve deep learning (DL) image reconstruction. The conventional protocol uses PAT 2 and an isotropic resolution of $1.0 \times 1.0 \times 1.0 \text{ mm}^3$ while the Deep Resolve protocol uses PAT 6 and an isotropic resolution of $0.8 \times 0.8 \times 0.8 \text{ mm}^3$. However, DL helps to reduce acquisition time by nearly 52%.

1A and **1D** show the initial orientation, while **1B**, **1C**, **1E**, and **1F** are multiplanar reconstructions (MPR).



2 T2w 3D FLAIR

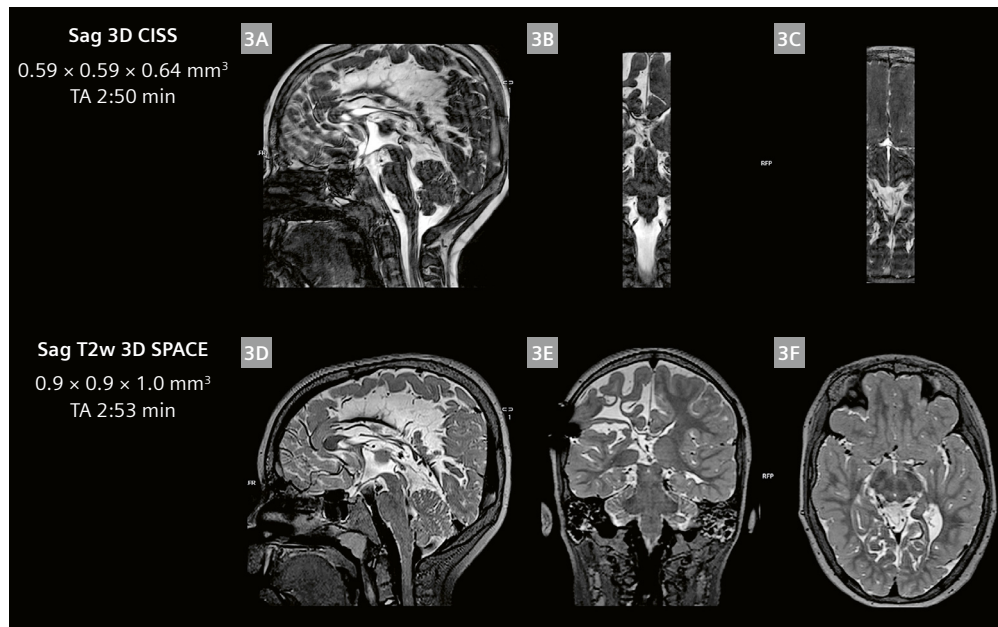
This figure demonstrates the difference between T2w 3D FLAIR images with and without Deep Resolve deep learning image reconstruction. The resolution and slice coverage are comparable in both sequences ($0.9 \times 0.9 \times 0.9 \text{ mm}^3$). The conventional protocol uses PAT 4, while the Deep Resolve protocol uses PAT 6. With Deep Resolve, it was possible to reduce the acquisition time by approximately 30% for this sequence. **2A** and **2D** show the initial orientation; **2B**, **2C**, **2E**, and **2F** show the corresponding MPRs.

Using Deep Resolve, it was possible to reduce the scan time by 52% and 30% for MPRAGE and 3D FLAIR, respectively, while improving the apparent image quality. While Deep Resolve Boost helped maintain a great signal-to-noise ratio (SNR) despite the higher acceleration factor, Deep Resolve Sharp helped improve the image quality by reconstructing

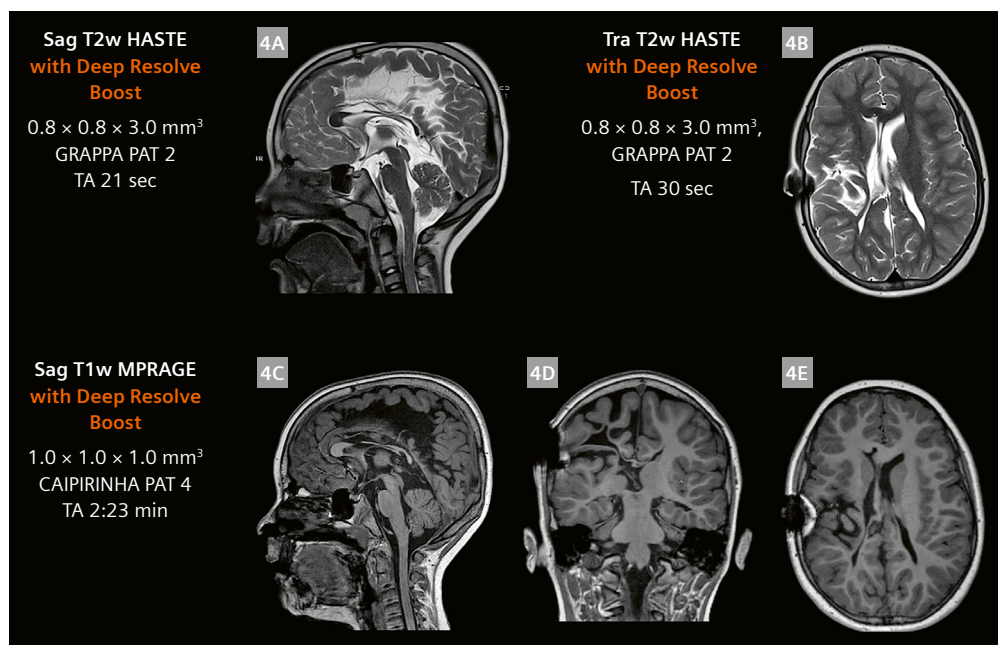
images with a super-resolution factor of 8. Since the resolution is improved in readout and phase-encoding direction by a factor of 2, we have a total factor of 4 in 2D imaging. In addition, with 3D Deep Resolve, you can add a factor of 2 in slice direction giving you an overall factor of 8.

Case 2

Awake MRI examination of a 7-year-old boy with a porencephalic cyst and difficulty remaining still. The patient is being monitored following a shunt placement.



3 The 3D CISS images (3A–3C) exhibit motion artifacts due to the patient's difficulty in remaining still. In contrast, the T2w 3D SPACE sequence (3D–3F) was successfully completed with good image quality. 3A, 3D initial sagittal orientation, 3B, 3C, 3E, and 3F corresponding coronal and transversal MPRs.

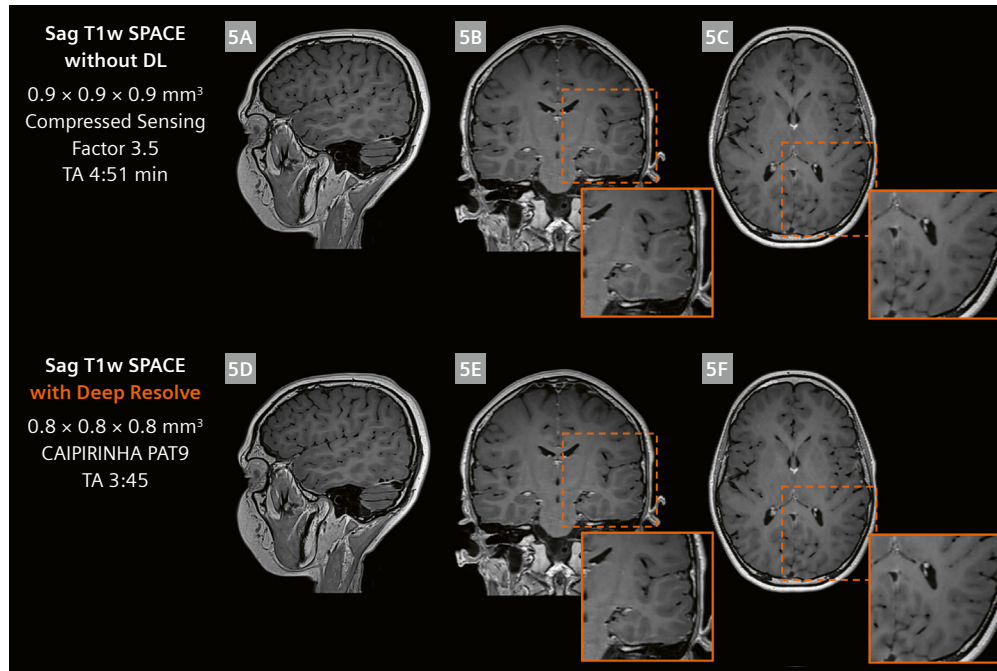


4 Continued from Figure 3. The boy expressed a desire to leave after the previous scan, but agreed to stay a few more minutes. We then performed a T2w HASTE sequence with Deep Resolve (4A and 4B), which allowed us to obtain sharp images in just a few seconds. Finally, we conducted the T1w MPAGE sequence with Deep Resolve (4C) as he reached his limit of tolerance. Figures 4D and 4E are MPR of 4C.

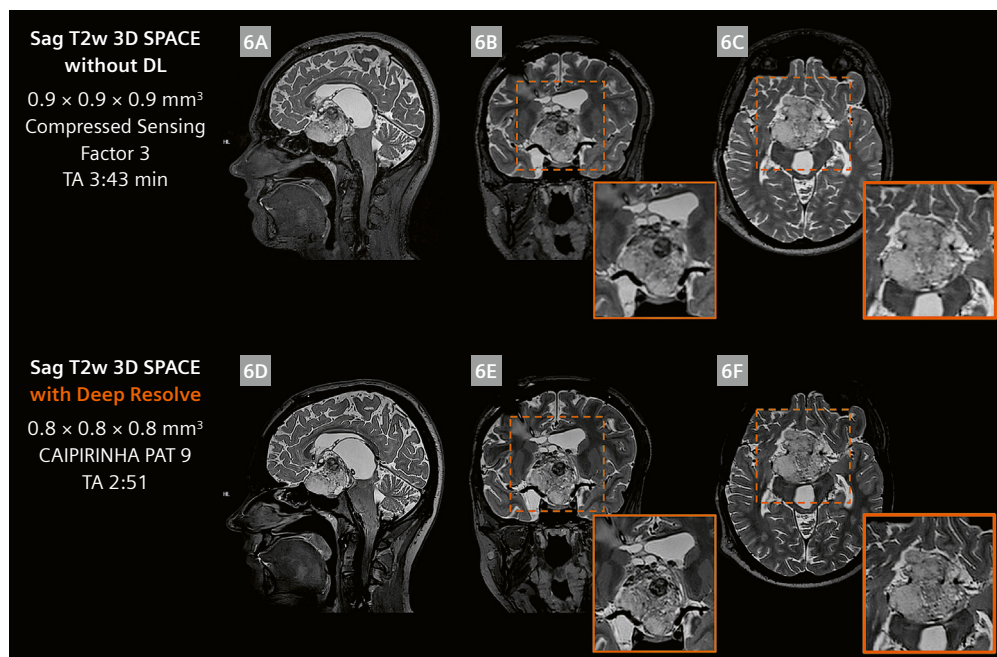
In this patient with limited cooperation, Deep Resolve was essential to performing a successful non-sedated MRI examination. Notably, without Deep Resolve, the routine T1w MPAGE sequence, which typically requires approximately 5 minutes, would likely have been unfeasible in this case.

Case 3

A 10-year-old boy underwent MRI without sedation for a known pilomyxoid astrocytoma in the suprasellar region.



5 This figure highlights a key challenge of 3D volumetric imaging: The MPR views often appear less sharp than the initial imaging orientations, which in this case is the sagittal orientation. Using the T1w 3D SPACE sequence, we compared routine acquisitions (5A–5C) with images acquired using Deep Resolve (5D–5F). The results clearly demonstrate improved sharpness and clarity of the MPR view with Deep Resolve. 5A and 5D initial sagittal orientation, 5B, 5C, 5E, and 5F coronal and transverse MPRs respectively.



6 Continued from Figure 5, we observe a similar effect with the T2w 3D SPACE sequence. The comparison between images without deep learning (6A–6C) and with Deep Resolve (6D–6F) reveals the same outcome: significantly sharper and clearer MPR views, without any signs of blurriness when applying Deep Resolve.

Here again, Deep Resolve allowed faster scans for both T1w and T2w SPACE while improving the image quality, especially when performing 3D reformatting.

Conclusion

The use of Deep Resolve 3D for pediatric neuro imaging shows great potential to further reduce the overall exam time beyond what was possible with Deep Resolve 2D. It furthermore contributes to improving image quality and the apparent resolution of the images. This may contribute to avoiding sedation or general anesthesia in some pediatric patients, while contributing to improve the diagnostic accuracy.

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Dr. Vossough is an adult and pediatric neuroradiologist in the Department of Radiology at Children's Hospital of Philadelphia. He holds the Larissa T Bilimuk Endowed Chair in Pediatric Neuroradiology Research.

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llVida_Fit_bacsiemens_pedimskknee_ultraflex_t8hlocalizer_dxt*
TA: 4 sec Coil Selection: Auto Voxel Size: 0.6x0.6x6.0 mm³ Acc: 2 Ref. SNR: 1.00

Properties
Start measurement without further preparation: Off
Wait for User to Start: Off
Start measurements: Off
Post-Process: Single Measurement
Auto Open Index Display: On
Auto Close Index Display: Off
Load Images to MRI ViewSQC: Off
Auto Show Images: On
Load Images to Stamp Segments: On
Load Images to Graphic Segments: On
Graphic: ungrouped: All Segments
Initial Mouse: Off

Resolution - Acceleration
Acceleration Factor PE: 2
Reference Lines PE: 24
Deep Resolve: Off
Phase Partial Fourier: Off
Sagittal Echo: Off

Resolution - Filter
Flow Filter: Off
Elliptical Filter: Off
POCS: Off
Distortion Correction: Off
Normalize: 2D
Image Filter: Thesson
Off

Routine
Scan Group: 1
Slices: 5
Distance Factor: 5
Position: Isocenter
Orientation: Transversal
Phase Encoding Dir: A >> P
Phase Oversampling: 0 %
FOV Head: 330 mm
FOV Phase: 100.0 %
Slice Thickness: 6.0 mm
TR: 7.0 ms
TE: 3.00 ms
Averages: 1
Contrastations: 5
AutoAlign: On

Contrast - Common
T1: 7.0 ms
T2: 3.00 ms
T2*: 0.00 ms
MTC: Off
Magn. Preparation: None
Flip Angle: 20 deg
Dark Band: Standard
Contrasts: 1
SSB: Off
Reconstruction: Off
Magnetics: Off

Contrast - Dynamic
Dynamic Mode: Standard
Measurements: 1
Multiple Series: Each Measurement

Resolution - Common
FOV Head: 330 mm
FOV Phase: 100.0 %
Slice Thickness: 6.0 mm
Noise Reduction: 246
Phase Resolution: 90 %
Interpolation: On

Resolution - Acceleration
Acceleration mode: GRAPPA
Reference Scans: Integrated

Geometry - Common
Scan Group: 1
Slices: 5
Distance Factor: 5
Position: Isocenter
Orientation: Transversal
Phase Encoding Dir: A >> P
Phase Oversampling: 0 %
FOV Head: 330 mm
FOV Phase: 100.0 %
Slice Thickness: 6.0 mm
TR: 7.0 ms
Multi-Slice Mode: Sequential
Interleaved: 5

Geometry - AutoAlign
Scan Group: 1
Position: Isocenter
Orientation: Transversal
Phase Encoding Dir: A >> P
Initial Position: Isocenter
H: 0.0 mm
V: 0.0 mm
Initial Orientation: Transversal
Initial Rotation: 0.00 deg

Geometry - Saturation
Saturation Mode: Standard
Spectral Saturation: None

Geometry - Tim Planning Suite
Switch-On Protocol: Off
Table Position: 0 mm
Table Rotation: 0 mm
Index Compensating: Off

System - Miscellaneous
Coil Selection: MSKA
Sagittal: S-C-T
Coronal: A >> P
Transversal: M >> S
Coil Combination: Additive Combine

Ultrashort-Echo-Time Magnetic Resonance Imaging for Pulmonary Evaluation in Children

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Abstract

Novel MRI techniques have been proposed to allow a radiation-free alternative to chest CT imaging to assess pulmonary anatomy in children. Ultrashort-echo-time (UTE) is a relatively new MRI technique that has the potential to provide high-resolution anatomical lung information. It also allows for the visualization of tissue structures with intrinsically short T2* components. We describe a feasibility study using a stack-of-spirals trajectory UTE sequence on a routine 1.5T MAGNETOM Avanto clinical MR scanner to depict lung structural findings in a series of children with and without pulmonary pathology.

Introduction

Chest CT is considered the standard imaging modality for lung evaluation in children. Although there is increasing scientific recognition that the perception of the risk of ionizing radiation from CT scanning may be greater than the true risk, this perception will likely continue to limit acceptance of serial CT scanning among families and physicians [1]. New MRI techniques for imaging the lung have been proposed as supporting a radiation-free alternative to CT imaging. These techniques are sensitive for the assessment and characterization of solid lung lesions and for the evaluation of infiltrative lung disorders [2], and could be of great utility in children at increased susceptibility to radiation, or in those who require numerous follow-up imaging studies.

In MRI, T2-weighted pulse sequences detect high or low signal from abnormal tissues. This strategy is useful for the diagnosis of parenchymal disease in most organs. However, some tissues, for instance in the lungs, have short T2s in which MR signal decays very rapidly and which

appear dark due to little or no signal detection when using echo times (TEs) from conventional clinical imaging [3]. Spin-echo imaging and fast gradient-echo sequences typically utilize a TE of approximately 8 milliseconds (ms), and 1 ms, respectively. However, radial and spiral sampling schemes have enabled the use of ultrashort-echo-time (UTE) and near-zero-echo-time (ZTE) methods, with TE values in the order of microseconds (μ s), which have allowed delineation of the short T2 components in these tissues [2]. Recently, there has been renewed interest in UTE techniques for imaging the lungs, which have an intrinsically short T2* relaxation time [4]. These sequences have been considered very valuable for pediatric lung imaging, due to their potential to provide high-resolution anatomical lung information [2].

In this technical report, we aim to describe the feasibility of using a 3D UTE sequence on a routine 1.5T MAGNETOM Avanto clinical MR scanner (Siemens Healthineers, Erlangen, Germany) to depict lung structural findings in children and to compare the imaging findings with the current standard chest CT imaging.

Methods

UTE acquisition sequence

Traditionally, most UTE acquisition sequences applied in musculoskeletal imaging for visualizing very short T2* structures, such as cartilage, use hybrid radial-based methods like pointwise encoding time reduction with radial acquisition (PETRA). In PETRA, the outer k-space is filled with radial half-projections, whereas the center of k-space is measured with a cartesian trajectory [5]. However (in addition to longer scan times due to the time needed for the cartesian imaging), in the presence of chemical shift,

motion, and B0 inhomogeneities, PETRA-based acquisitions suffer from radial phase shift artifacts that make it unsuitable for chest imaging.

The acquisition sequence in this study uses a stack-of-spirals trajectory, which ensures motion-robustness and reduces the minimum time required to cover k-space compared to a 3D radial trajectory (Fig. 1A) [6, 7]. The sequence is based on variable-duration slice encoding to minimize T2* decay, on cartesian sampling along the through-plane direction, and on in-plane spiral sampling to reduce the minimum time needed to cover the entire k-space. The acquisition sequence uses a non-selective radiofrequency (RF) pulse, which excites all tissue in the scan region. Therefore, we acquire the data in coronal orientation with isotropic voxels to minimize the number of through-plane phase encoding steps. The isotropic nature of the acquisition allows the data to be viewed in multiple orientations. It is retrospectively reformatted to allow a better view of features of interest, and to generate views that correspond to slices acquired with other sequences, if desired. A sequence diagram can be seen in Figure 1B. Each spiral readout starts in the center of k-space. The TE

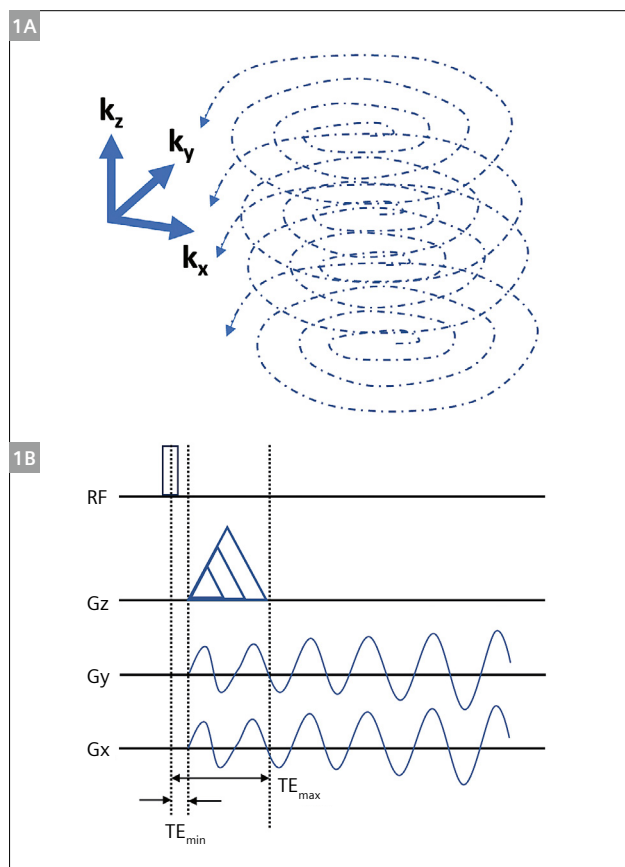
is defined as the time between the center of the RF pulse and the acquisition of the k-space center. To minimize the echo time, a short non-selective excitation pulse is used. In addition to minimizing the echo time itself, the non-selectivity saves time for the slice refocusing gradient. Slice encoding is performed with a phase encoding gradient between excitation and spiral readout. The time required for the 3D phase encoding gradient increases with distance from the center partition. For the central partition, no gradient is needed at all. Each spiral readout starts immediately after the 3D phase encoding gradient. The central k-space partition determines the overall contrast of the image, and hence the protocol TE is equal to the minimum TE.

Patients

The Institutional Review Board approved this prospective study for sequence development and optimization on the clinical scanners. We included patients who were scheduled for chest or abdominal MRI as part of their clinical care and when, in most cases, a concurrent chest CT scan was also required for clinical purposes. A chest UTE MRI sequence was successfully performed following completion of the routine clinical sequences on 9 children (6 boys, 3 girls) ranging in age from 1¹ to 17 years (median 9 years). 8 patients were scheduled for an abdominal MRI, and one was scheduled for a chest MRI. All of these patients underwent a clinical chest CT scan on the same day as their MRI. The clinical indications for imaging were rhabdomyosarcoma (n = 1), sarcoma of the liver (n = 1), Wilms tumor (n = 2), Fanconi anemia (n = 1), yolk sac tumor with lung metastases (n = 1), Ewing's sarcoma (n = 1), osteosarcoma (n = 1), and soft tissue sarcoma (n = 1).

CT imaging

All patients underwent CT imaging on the same day as their MRI, as part of their clinical care. Chest CT examinations were performed on a 128-slice multidetector SOMATOM Definition CT scanner (Siemens Healthineers, Forchheim, Germany) with the following settings: 128 × 0.6 mm slice collimation; 0.28 s gantry rotation time; 75 ms temporal resolution. The automated tube voltage and current were set according to patient size, and a weight-based CT technique was used to minimize radiation dose. A single-phase acquisition was performed during inspiration. Images were reconstructed in axial, coronal, and sagittal planes. Six patients had chest CT imaging without contrast, and two patients were scanned following the intravenous administration of Omnipaque 300.



1 (1A) Stack-of-spirals trajectory and (1B) sequence diagram of the UTE spiral acquisition.

¹MR scanning has not been established as safe for imaging fetuses and infants less than two years of age. The responsible physician must evaluate the benefits of the MR examination compared to those of other imaging procedures.

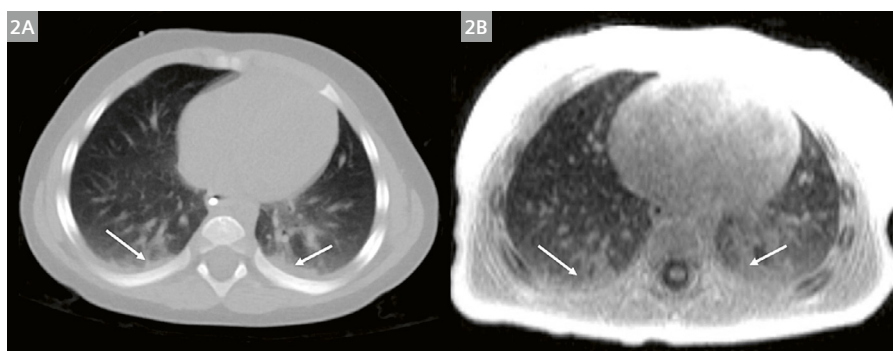
MR imaging

All patients were scanned on a 1.5T MAGNETOM Avanto clinical scanner. Two patients were sedated on free breathing for the procedure, and one infant received general anesthesia for the scan. For all studies, the built-in body volume coil was used for signal intensity transmission, and a standard 8-channel chest coil was used for signal intensity reception. A 3D isotropic UTE imaging sequence² (investigational work-in-progress) was acquired in the coronal plane, and axial and sagittal reformats were generated. An 8-bit grayscale filter was applied to the axial reformat images, improving the contrast between the more hypointense lung parenchyma and the hyperintense

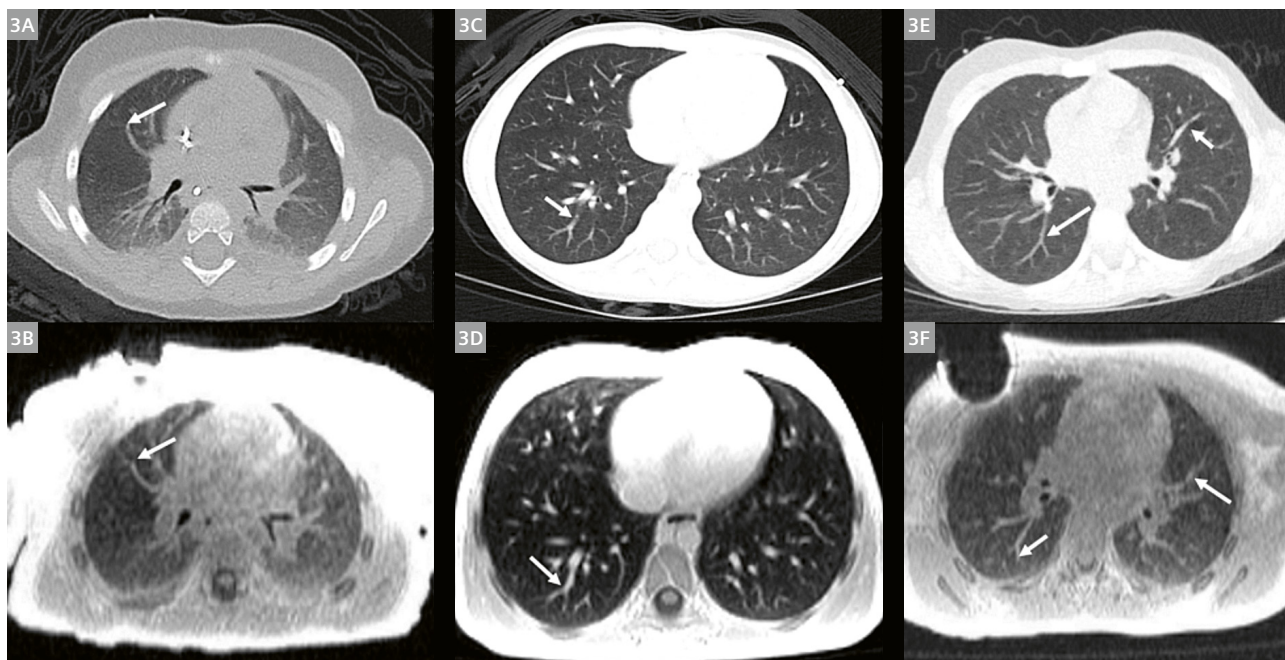
interstitium, when compared to the contrast without a filter, resulting in a visual experience more closely resembling that of reading a CT scan. The typical parameters used are described in Table 1.

The images obtained at the TE of 50 μ s were defined as the UTE images. Acquisition was performed during free-breathing. Prospective gating was used to reduce respiratory-related artifacts: navigator pulses were inserted into the sequence, from which the respiratory signal was extracted; once the sufficient spiral data was collected during end-expiratory state (acceptance window: 40%), the scan stopped automatically. Therefore the total scan time depends on patient's respiratory pattern, and was approximately 4 minutes per acquisition, a duration fairly similar to that of other conventional pulse sequences. The signal acquired with the UTE sequence is only from

²Work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.



2 A 1-year-old boy¹ with a history of rhabdomyosarcoma of the pelvis, post-chemotherapy and radiation treatment, with evidence of bilateral atelectasis in the lower lobes (arrows), similarly depicted on chest CT imaging (2A) and a UTE sequence in MRI (2B). A 3 mm right middle lobe nodule detected on both CT.



3 Comparison of axial chest CT imaging findings and a corresponding axial UTE sequence in chest MRI in children of varying ages: A 1-year-old boy¹ with a diagnosis of rhabdomyosarcoma (3A, 3B), and a 7-year-old girl (3C, 3D) and a 4-year-old girl (3E, 3F), both with a diagnosis of Wilms tumor. All cases demonstrate adequate visualization of peripheral and central vessels (arrows) on UTE imaging.

structures with very short T2 relaxation times. One thoracic pediatric radiologist (JBR) with five years of experience, reviewed the chest CT images and compared them to the UTE MR images. Atelectasis was identified in two patients, visualized similarly on both UTE and CT imaging (Fig. 2). Three patients had sub-centimeter nodules. UTE was able to resolve nodules of 3 mm and larger (Fig. 2). A single 2 mm solid nodule in one patient was seen only on CT imaging. Central and peripheral pulmonary vessels were similarly visualized by both techniques (Fig. 3) as was bronchiectasis (Fig. 4). In one patient, a dilated peripheral upper lobe bronchus was not visualized with chest UTE MRI (Fig. 5).

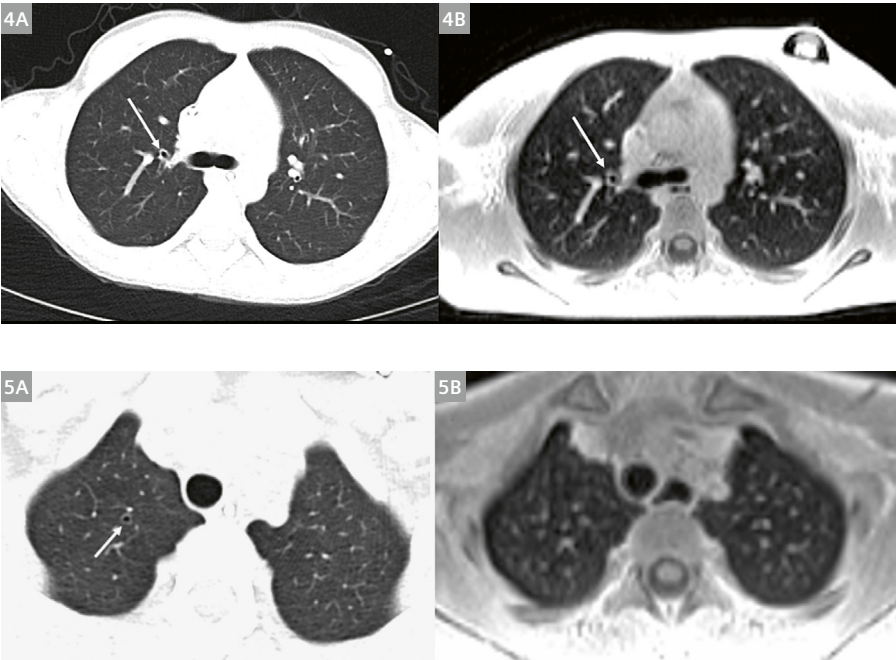
Results

We report on a novel spiral 3D isotropic UTE MRI technique² in the evaluation of lung diseases in pediatric patients. In this UTE MRI feasibility study, we compared UTE MR images in children with and without lung disorders to their respective chest CT images. Our comparison demonstrated that most features of lung disease detected by the CT scans were also detected by UTE MRI.

Parameter	UTE MRI
Pulse sequence type	3D
Matrix	320 × 320
No. of signals acquired	1
TR/TE	3.7 ms / 50 μs
No. of slices	150–200
Slice thickness (mm)	1
Gap (mm)	0
FOV setting (mm)	300
Flip angle (°)	5
Spatial resolution (mm)	1.0
Approximate scanning time	4–5 min

Table 1: UTE MRI acquisition protocol (Adapted from [2])

TE = echo time,
TR = repetition time,
FOV = field of view,
mm = millimeters, min = minutes,
μs = microseconds,
ms = milliseconds,
No. = number



- 4

A 6-year-old girl with a history of Wilms tumor status post resection and radiation treatment, with evidence of right upper lobe bronchiectasis (arrow) on axial chest CT imaging (4A) and on an axial UTE sequence in chest MRI (4B).
- 5

A 6-year-old girl with a history of Wilms tumor status post resection and radiation treatment, with evidence of a solitary dilated peripheral bronchus in the right upper lobe (arrow) on chest CT imaging (5A), not visualized on the UTE sequence on chest MRI (5B).

Discussion

CT is the most established lung imaging modality because of its ability to localize and quantify the findings of lung diseases in children. However, CT imaging exposes children to ionizing radiation, which poses undesirable concerns for patients, families, and physicians, especially in lifelong longitudinal assessments of disease progression. MRI does not involve ionizing radiation, which mitigates the risk of future radiation-induced conditions, and can provide valuable functional and anatomical information.

Hyperintense or hypointense signal on T2-weighted images depends on the T2 relaxation times of the organs being imaged, and these differ depending on the proton density in the tissues. The short effective transverse relaxation time of the lung parenchyma results in a loss of image quality and compromises the delineation of pulmonary anatomical information. MRI studies using conventional T1- and T2-weighted sequences have found MRI to be inferior to CT imaging in pediatric lung disease evaluation. New UTE-based MRI techniques can bridge this gap in the quality of anatomical and functional lung evaluation, which is not currently possible by other means. Novel methods based on UTE imaging have improved the lung image quality, with imaging findings approaching those of a chest CT scan. The technique achieves these improvements by reducing the TE to 100 μ s or less, minimizing the parenchymal signal decay caused by short effective transverse relaxation time. Therefore, these methods can detect very subtle lung physiology, including gravity-dependent lung-density gradients [8, 9].

There is a great need for a common methodology that can detect and monitor early signs of morphological and functional lung changes in patients with lung disorders from birth throughout their life. Thus, UTE MRI may be integrated into routine clinical MRI sequences, along with contrast-enhanced T1-weighted imaging and perfusion imaging, to provide a more comprehensive evaluation of pediatric lung diseases.

Conclusions

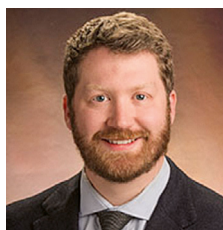
The UTE MRI sequence is advantageous for imaging the lungs in children because it can be acquired during free breathing. We conducted a feasibility study for a stack-of-spirals-trajectory UTE sequence on a standard 1.5T clinical MRI scanner to visualize lung structures in a group of children, both with and without pulmonary pathology.

Acknowledgment

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Phase-Resolved Functional Lung (PREFUL) MRI: The Next-Generation Ventilation-Perfusion Scan?

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Introduction

Chronic pulmonary diseases – such as chronic obstructive pulmonary disease (COPD), cystic fibrosis (CF), and chronic thromboembolic pulmonary hypertension (CTEPH) – require accurate and sensitive diagnostic techniques to monitor progression and therapeutic response [1]. Traditional pulmonary function tests (PFTs), including the Tiffeneau-Pinelli index, provide global lung function parameters but lack the capacity to map region-specific changes [2]. Imaging approaches like computed tomography (CT) or single photon emission computed tomography (SPECT) can visualize lung morphology and function but inevitably involve ionizing radiation, making repeated examinations challenging [3, 4].

Magnetic resonance imaging (MRI) offers a promising, radiation-free alternative for lung function assessment. Although low proton density and rapid signal decay in the lung pose technical challenges [5], several methods have been developed to address these limitations. For instance, fluorinated ¹⁹F or hyperpolarized gas MRI (e.g., ¹²⁹Xe) provides direct ventilation imaging [6] and, when paired with intravenous contrast [7], yields perfusion information. However, such techniques require specialized equipment, trained personnel, and contrast agents, and they may raise concerns over long-term safety.

Fourier decomposition (FD) is a contrast-free approach that uses conventional proton MRI and postprocessing to derive ventilation (V) and perfusion (Q) maps [8]. After image registration, the respiratory and cardiac frequency components of the time-series signal in each voxel are extracted, enabling simultaneous V/Q assessment. Soon, advanced versions of FD were developed [9–11],

including self-gated non-contrast-enhanced functional lung imaging (SENCEFUL) [12]. Incorporating an additional DC acquisition in a quasi-random fast low-angle shot (FLASH) sequence allowed to sort each individual *k*-space line into one synthetic respiratory and cardiac cycle for higher temporal resolution and analysis of ventilation perfusion dynamics.

Following this idea, phase-resolved functional lung (PREFUL) MRI uses extended post-processing, allowing for retrospective image sorting according to respiratory and cardiac phases in combination with a conventional unmodified spoiled gradient-echo (GRE) or balanced steady state free precession (bSSFP) acquired in free breathing [13]. This allows dynamic V/Q assessment with commercially available sequences. Consequently, PREFUL MRI permits region-specific evaluations of dynamic lung function, including ventilation, perfusion, and even phase-based parameters such as flow-volume loop correlation metric (FVL-CM) [14] and pulse wave transit time or velocity measurements [15, 16]. Furthermore, many clinical studies have shown robust correlations with conventional methods [17–19, 15], reproducibility across different centers [20–22], and sensitivity in therapeutic and monitoring situations [23–29].

Despite these promising results, PREFUL remains restricted to a few specialized centers, partly due to limited awareness of the method's technical feasibility and clinical potential. This work aims to provide an overview of the current PREFUL methodology, thereby facilitating broader implementation.

Theory

Four main components – proton density, time-of-flight (TOF) effects, movement, and noise – primarily influence the pulmonary time-series MR signal in free-breathing spoiled gradient-echo acquisitions. Additional factors (e.g., T1, T2/T2* decay, diffusion, field inhomogeneities) are typically minimized through appropriate protocols (low flip angle, short TR, asymmetric echo, and dynamic parallel imaging [30]) and thus are omitted here. Pulmonary and cardiac motion during the MR acquisition is compensated by image registration. During expiration, shrinking lung volume increases proton density and thus raises the MR signal; during inspiration, the opposite occurs, forming the basis of ventilation measurements through the signal-volume relationship [31, 32]:

Equation 1

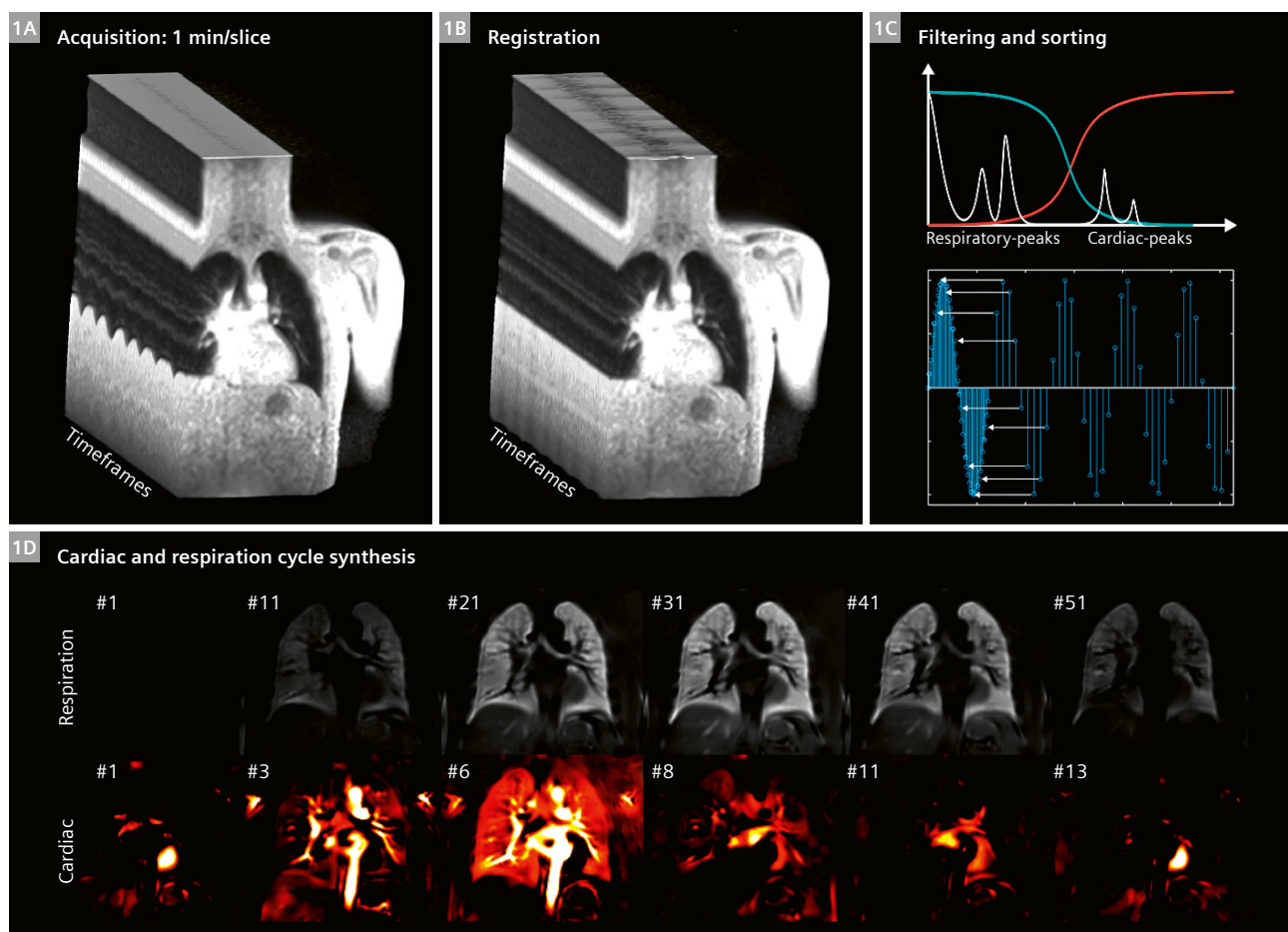
$$S \sim 1 / V$$

Pulsatile blood inflow induces further signal modulations known as TOF effect, which underpin perfusion measurements. Given the lung's inherently low signal, averaging and filtering improve signal-to-noise ratio so that the remaining variations predominantly reflect ventilation and perfusion. Because these variations occupy distinct frequency ranges, they can be separated by Fourier analysis [33]. Ventilation is then quantified via the inverse relationship between lung volume and proton signal, yielding the regional ventilation (RVent) [34] parameter:

Equation 2

$$RVent = \frac{V_{Insp} - V_{Exp}}{V_{Reg}} = \frac{S_{Reg}}{S_{Insp}} - \frac{S_{Reg}}{S_{Exp}}$$

During the initial transient phase of the acquisition, the MR signal has not yet reached steady state (SS) and thus briefly reflects near-maximal magnetization. By fitting these early



1 Simplified illustration of the phase-resolved functional lung (PREFUL)¹ MRI workflow: **(1A)** Free-breathing spoiled gradient-echo acquisition, **(1B)** motion correction via image registration, **(1C)** signal filtering (high-pass (red curve) and low-pass (blue curve)) and phase sorting, and **(1D)** synthesis of the cardiac and respiratory cycles (subset of phases provided). The resulting data are then quantified to yield various functional parameters and generate a final report.

data points to an exponential curve, it is possible to estimate the maximum signal (S_0). In addition, the signal drop between S_0 and steady state captures the potential effect if all voxels were moving. Together, these measurements serve as normalization factors for blood fraction (BF) content:

$$\text{Equation 3} \quad \text{BF} = S_{0,\text{Map}} / S_{0,\text{Vessel}}$$

and for flow-related signal difference Q in the steady state:

$$\text{Equation 4} \quad \text{EF} = Q / \text{Median}(S_0 - \text{SS})$$

as fully detailed by Glandorf et al. [35], and allow full quantification of perfusion (QQ) in mL/min/100 mL:

$$\text{Equation 5} \quad \text{QQ} = \text{EF} \cdot \text{BF} \cdot \text{VV} \cdot f_{\text{Heart}} \cdot \frac{60 \text{ s}}{\text{min}} \cdot \frac{100}{\text{VV} \cdot 100}$$

with the heart frequency f_{Heart} determined from the fourier spectrum and the voxel volume (VV).

Methods

Figure 1 gives a simplified overview of the core protocol steps, consisting of acquisition, registration, filtering and sorting, and final cycle synthesis. Further details are described in the following sections. More in-depth information can be found in previous articles [13, 36].

For this article, retrospective evaluation of unpublished data from a healthy volunteer and a representative CF patient who participated in a previously published study on elexacaftor-tezacaftor-ivacaftor (ETI) therapy [28] was performed.

Acquisition

After standard MRI safety precautions and prescreening for contraindications, participants are placed head-first and supine on a 0.55T, 1.5T, or 3T scanner table. A multi-channel flex coil is positioned beneath the chin to

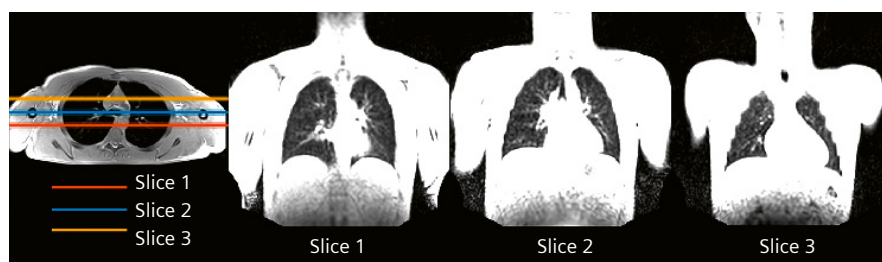
optimize signal coverage of the lungs, and care is taken to secure it without restricting breathing. Localizer scans establish orientation, and a transversal morphological image is acquired to identify the tracheal bifurcation, which serves as a consistent landmark for coronal slices. Depending on the protocol length, either a limited number of slices (e.g., three slices spaced at 100%) or multiple slices with smaller spacing (20%–33%) are acquired to cover the lungs; each slice is acquired separately rather than interleaved. Typical temporal resolutions range from approximately 3 to 5 images/second. A representative protocol is provided in Table 1. Example slice positioning is shown in Figure 2.

Parameter	Setting
Sequence	SPGRE, bSSFP
Orientation	Coronal
Field of view (FOV)	500 × 500 mm ²
Slice thickness (ST)	15 mm
Repetition time (TR)	3 ms or lower
Echo time (TE)	1 ms or lower
Bandwidth (BW)	1500 Hz/Pixel
Flip angle (FA)	~5° (Ernst angle)
Acceleration	T-GRAPPA 2×
Temporal resolution (dt)	> 3 frames/second
Frames/measurements	up to 500 depending on dt
Total duration / slice	1 minute

Table 1: Typical sequence settings for a PREFUL¹ acquisition. Please note that some settings such as TE, TR, and FA will depend on the field strength and gradient system.

Postprocessing

Following acquisition, images are corrected for respiratory and cardiac motion, filtered for noise reduction and V/Q separation, and sorted to generate phase-resolved data for both ventilation and perfusion.



2 Example of slice positioning using a transverse volumetric interpolated breath-hold examination (VIBE) sequence for planning. Then, three coronal slices are acquired for PREFUL in free breathing with a 100% slice gap. The second slice is aligned with the tracheal bifurcation to serve as a repeatable anatomical landmark.

General segmentation

To enable a fully automated processing pipeline, all necessary segmentations are performed by either neural networks (nnU-Net/U-Net for lung boundaries [37] and vessel segmentation [38, 16]) or algorithmic methods for the large central vessel [18]. The final lung parenchyma region of interest (ROI) is then generated by subtracting the vessel segmentation from the lung boundary ROI. Figure 3 provides an illustration of the segmentation procedure.

Registration

Images are grouped into ten respiration-based categories (inspiration to expiration). Non-rigid registration is performed within each group toward an intermediate lung position, then progressively between groups until they converge to a mid-respiration reference (Group 5). This approach (GOREG) minimizes the deformation per registration step and enhances algorithmic stability [39]. As registration packages Advanced Normalization Tools (ANTs) [40] with BSplineSyN is known to deliver a good registration quality. However, the less-explored polynomial expansion-based algorithm ("Forsberg") [41, 42] can offer comparable results with shorter processing times [43, 44].

General filtering

Registered images are denoised using guided filtering [45] with the temporally averaged image as a guiding image. Low-pass and high-pass filters (cut-off around 0.7 Hz) are applied to separate perfusion- and ventilation-related components. The initial 20 images are discarded (except in quantified perfusion calculations) to ensure a steady-state signal.

Perfusion

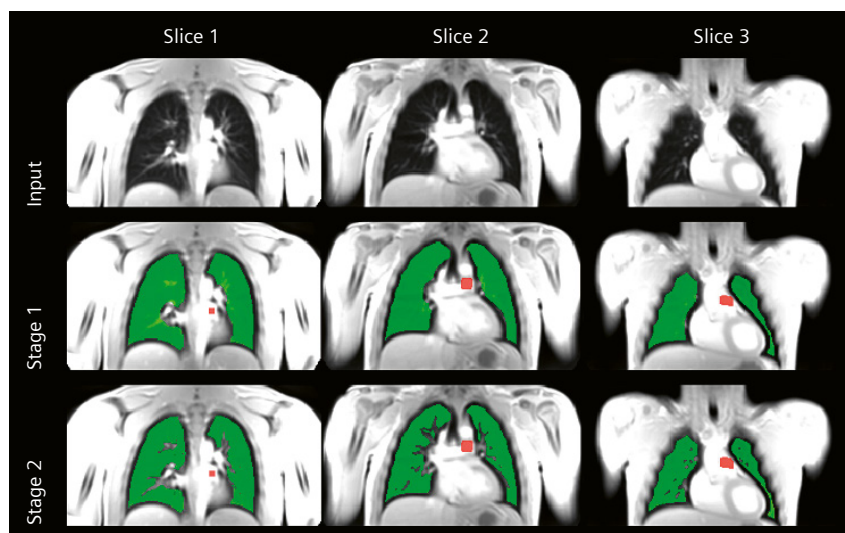
Cardiac phase determination relies on algorithmically identifying strong perfusion-weighted signal regions in the lung and mediastinum (red ROI, Fig. 3) [18]. A piecewise sinusoidal fit of the corresponding time-series allows phase-sorting of the cardiac cycle, which is then interpolated (15 phases) to exceed the original sampling rate and generate high-temporal-resolution perfusion dynamics. This re-sorted dataset provides an entire cycle from diastole to systole, and back to diastole (Fig. 1).

Ventilation

Respiratory frequency and tidal volume can vary substantially, so a cosine model function is used to assign images to specific respiratory phases based on signal amplitude. The slope of the signal determines whether data belongs to inspiration or expiration. Nadaraya-Watson kernel regression then generates a uniformly sampled respiratory cycle. RVent is calculated by comparing the mid-respiration signal with inspiration and expiration signals (Equation 2) [34].

Parameter calculation and thresholding

The synthesized full cardiac and respiratory cycles enable derivation of additional functional parameters. RVent is computed at each phase, while FVL-CMs are calculated as an MRI analog to flow-volume loops of pulmonary function tests [14]. Quantified perfusion involves normalizing early transient-state images to account for proton density changes, fitting an exponential model to reach maximal magnetization, and then identifying cardiac phases with maximal lung parenchyma signal [46, 35]. Blood fraction and exchange fraction estimates are combined to calculate perfusion in mL/min/100 mL (Equation 5).



3 Illustration of the segmentation procedure for three slices. In the first stage, a neural network identifies the lung boundary (green). In the second stage, intraparenchymal vessels are excluded from the lung boundary. Separately, an algorithm segments a large vessel with a strong perfusion signal (red).

Parameter maps undergo static and dynamic thresholding to classify ventilation and perfusion defects (VD/QD). FVL-CM is thresholded with a fixed value of 90%. For the remaining parameters, the 90th percentile is used as an estimate for a normal value and multiplied with an empirical factor (0.4 for RVent and 0.15 for QQ).

Corresponding defect percentages are computed based on the proportion of voxels below the chosen thresholds relative to total lung parenchyma. Combining RVent and FVL-CM allows further subcategorization of V/Q defects according to a fourfold table (see Table 2).

Defect	VD	No VD
QD	VQM(D)	QDP(Ex)
No QD	VDP(Ex)	VQM(H)

Table 2: The resulting four combinations when comparing ventilation and perfusion defect (VD/QD): VQ match of defects (VQM(D)), VQ match of healthy regions (VQM(H)), and exclusive VDP and QDP.

All analyses in the results section use Forsberg registration and combine both RVent and FVL-CM metrics into a single VD measurement, classifying a voxel as VD if either RVent or FVL-CM (or both) met the defect criteria. Lung parenchyma parameters are summarized using mean values, and the complete set of ventilation/perfusion (V/Q) metrics is reported.

Results

Example data from a healthy volunteer (Fig. 4) demonstrate homogeneous ventilation and perfusion throughout the lung parenchyma, resulting in a 92% V/Q match. By contrast, a cystic fibrosis patient (Fig. 5) shows heterogeneous distribution before treatment, corresponding to a 40% V/Q match. Following therapy, all parameters improve markedly, culminating in a 93% V/Q match.

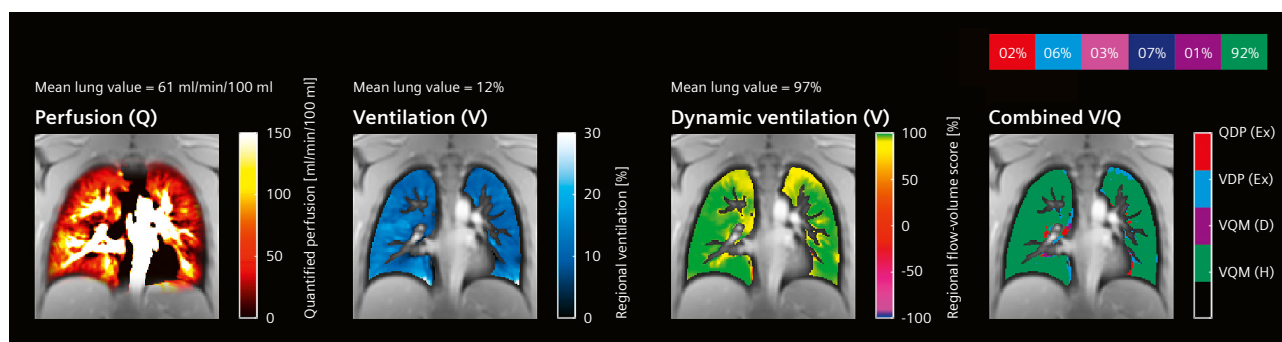
Discussion

In this article, we described the proton MRI technique known as PREFUL, providing example results to highlight its capabilities. A key distinction of PREFUL, compared to techniques like Fourier decomposition and its derivatives,

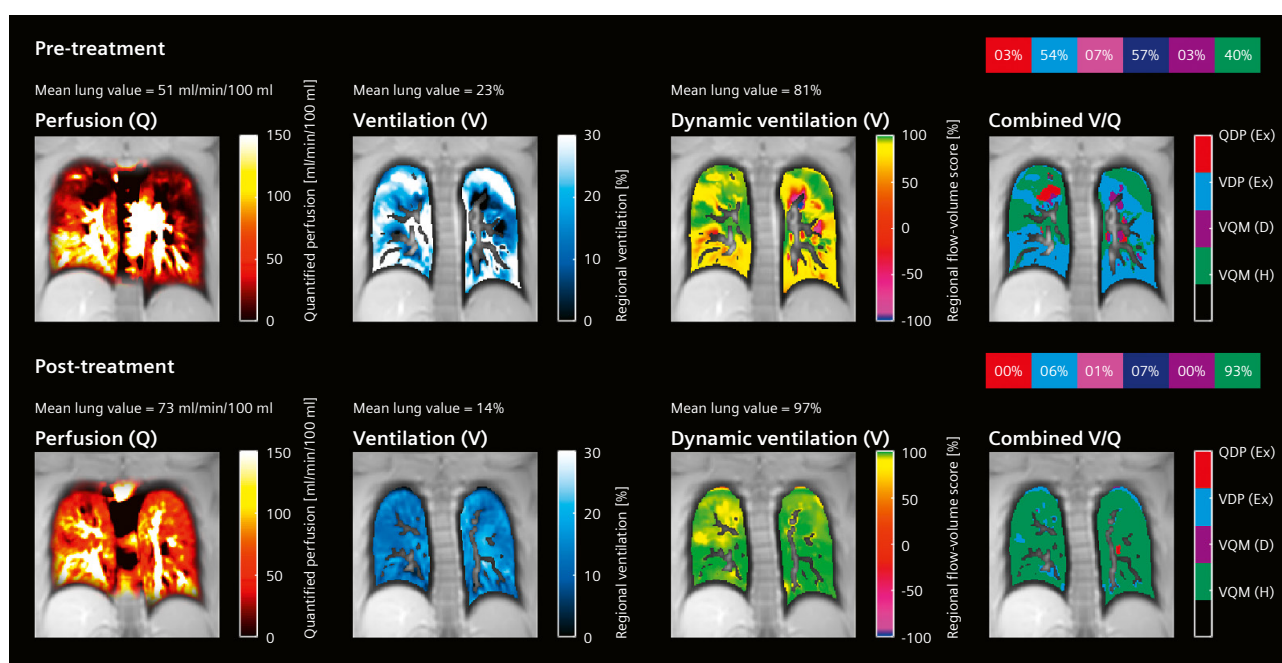
is its combination with a conventional spoiled gradient echo (SPGRE) sequence. While balanced steady-state free precession (bSSFP) sequences deliver a signal-to-noise ratio advantage at 1.5T or lower, at 3T they suffer from reduced image quality and banding artifacts that hinder analysis [47]. Despite being developed for SPGRE, recent studies have demonstrated that PREFUL provides strong comparability with both sequences at 1.5T [48]. Besides the image sorting algorithm, as an additional major difference in post-processing, filtering is less restricted and amplitude calculation is carried out in time domain to account for frequency variations.

As outlined in the introduction, PREFUL has undergone validation against gold standards like SPECT, dynamic contrast-enhanced (DCE) MRI, and (hyperpolarized) gas MRI [49, 17, 18, 50, 19]. It has been tested in a variety of settings [51, 52, 20, 48, 44], including multi-center trials [21, 22, 53] and at 0.55T [25, 26, 54], demonstrating robust performance. PREFUL has proven capable of reflecting disease pathology and tracking disease progression and treatment responses [23, 55, 56, 24, 27, 57, 28, 16], making it a promising candidate for clinical translation. In particular, dynamic parameters like the spirometry-inspired FVL-CM have shown greater sensitivity as biomarkers compared to static measures that rely on only a limited portion of the synthesized cycles.

The main limitation of the PREFUL method lies in its coarse spatial resolution ($4 \times 4 \times 15 \text{ mm}^3$) and moderate acquisition time (1 minute per 2D slice), which complicate detailed lung assessments. To address this, a 3D version of PREFUL has been developed, offering higher isotropic resolution ($4 \times 4 \times 4 \text{ mm}^3$) and enabling more comprehensive ventilation dynamics measurements with a self-gated stack-of-stars VIBE or FLORET UTE sequence [58, 59] in about eight minutes. Despite being a relatively new technique, several studies have already validated the clinical applicability of 3D PREFUL [60, 43, 61]. However, a primary drawback of the 3D version is the lack of integrated perfusion information and the need for more specialized sequences. Consequently, 2D PREFUL remains a forward-looking technique, particularly as emerging biomarkers – such as pulse wave transit time or velocity measurements – gain traction. Ongoing advancements in image acceleration and protocol refinements may offer a near-term pathway to broader dissemination of the 2D PREFUL method.



4 Example PREFUL maps of a healthy volunteer (male, age 36), showing perfusion in mL/min/100 mL, regional ventilation in %, flow-volume loop correlation metric (FVL-CM) in %, and a V/Q map obtained by thresholding and combining all maps. Please note the homogenous distribution of V/Q values in the lung parenchyma.



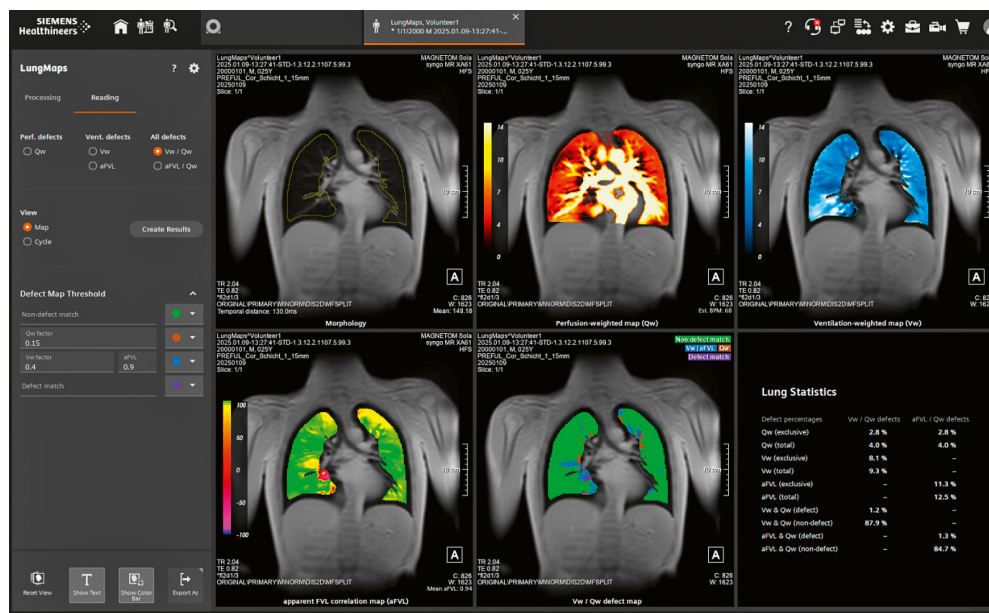
5 Pre- and post-treatment effect visualized for a CF patient (male, age 20) after ETI therapy. Please note the homogenization of V/Q after treatment, visible on all maps.

The LungMaps application from Siemens Healthineers

PREFUL technology will be available soon under the product name LungMaps¹ as a postprocessing application for MRI scanners from Siemens Healthineers. However, the nomenclature in the product is slightly different:

- Rvent is called "Ventilation-weighted map" (Vw) without quantitative numbers.
- QQ is called "Perfusion-weighted map" (Qw) without quantitative numbers.
- FVL-CM is called "apparent flow volume loop correlation map" (aFVL).
- In the defect maps (see Table 2 and Figure 6), the product uses the following labels:
 - "Defect match" instead of VQM(D)
 - "Ventilation defect" instead of VDP(Ex)
 - "Perfusion defect" instead of QDP(Ex)
 - "No defect match" instead of VQM(H)

¹ LungMaps is currently under development and not commercially available. Its future availability cannot be ensured.



6 The LungMaps¹ application from Siemens Healthineers. This dataset from a healthy volunteer shows regular perfusion- and ventilation-weighted maps. The aFVL map and the Vw/Qw defect map show no signs of lung defects. The lower right corner displays a structured report of the defect percentages, which is also available as a PDF and as a secondary capture DICOM file.



To learn more about LungMaps please visit us at

siemens-healthineers.com/magnetic-resonance-imaging/options-and-upgrades/clinical-applications/lungmaps

Figure 6 shows a screenshot of the application running on *syngo.via*, in this case on a healthy volunteer. All processing, including the neural network-based segmentation (see General segmentation) of the lungs and PREFUL analysis, is fully automated. However, the user can manually edit the lung segmentation if desired

Conclusion

PREFUL MRI provides a robust, radiation-free means of assessing ventilation and perfusion in a single free-breathing scan. Validation studies have consistently shown strong concordance with established methods like SPECT, DCE, and hyperpolarized gas MRI, underscoring its clinical potential. Although current 2D acquisitions use relatively coarse spatial resolution, the 3D version provides more detailed lung coverage, albeit without simultaneous perfusion and with less widespread sequences. Ongoing refinements in acquisition speed, resolution, and biomarker integration thus position 2D PREFUL as a promising candidate for next-generation V/Q scans using free-breathing proton-based pulmonary imaging.

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Myocardial Tuberculosis: A Curable Form of Cardiomyopathy in Endemic Countries. Know This Great Masquerader

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Abstract

Pericardial affection in tuberculosis has been widely described in the literature. Myocardial involvement is considered a rare manifestation, but is associated with high mortality. Variations in clinical and imaging presentations can often cause diagnostic dilemmas and further delay appropriate management. We share our experience of myocardial involvement in histopathologically proven cases of tuberculosis. The main intention is to create awareness about this rare but curable form of cardiomyopathy.

Introduction

Tuberculous infection is widespread in tropical countries. It occurs due to the spread of mycobacterium tuberculosis. The lungs are primarily involved. However, extrapulmonary involvement (lymph node, central nervous system, bone, cutaneous, abdominal) is also not uncommon (10%–15%) [1].

Cardiovascular involvement is rare and accounts for 1%–2% of extrapulmonary tuberculosis [2]. It is often associated with high mortality and morbidity. Pericardium is most commonly involved [3]. This can

present as pericardial effusion, constrictive pericarditis, and pericardial calcification.

Myocardial tuberculosis is a rare but potentially life-threatening form of tuberculosis. A high index of suspicion and multimodality imaging workup is needed for evaluation. Confirmation of diagnosis with histopathology workup is essential to improve outcomes with prompt anti-tubercular treatment.

Tuberculosis can reach the myocardium through:

1. Hematogenous spread from primary infection or reactivation of latent focus.
2. Direct extension from adjacent infected structures like the pericardium and mediastinal lymph nodes.
3. Lymphatic spread through infected lymph nodes.

Clinical presentation is highly variable depending on the severity and location of myocardial involvement. Systemic symptoms like fever or weight loss may not be present. Patients can manifest with rhythm disturbances, cardiac failure, dilated cardiomyopathy, or (rarely) sudden cardiac death.

This article describes interesting cases of myocardial involvement in tuberculosis. Images were acquired using our 3T MAGNETOM Verio and 1.5T MAGNETOM Altea scanners (Siemens Healthineers, Erlangen, Germany).

Case 1

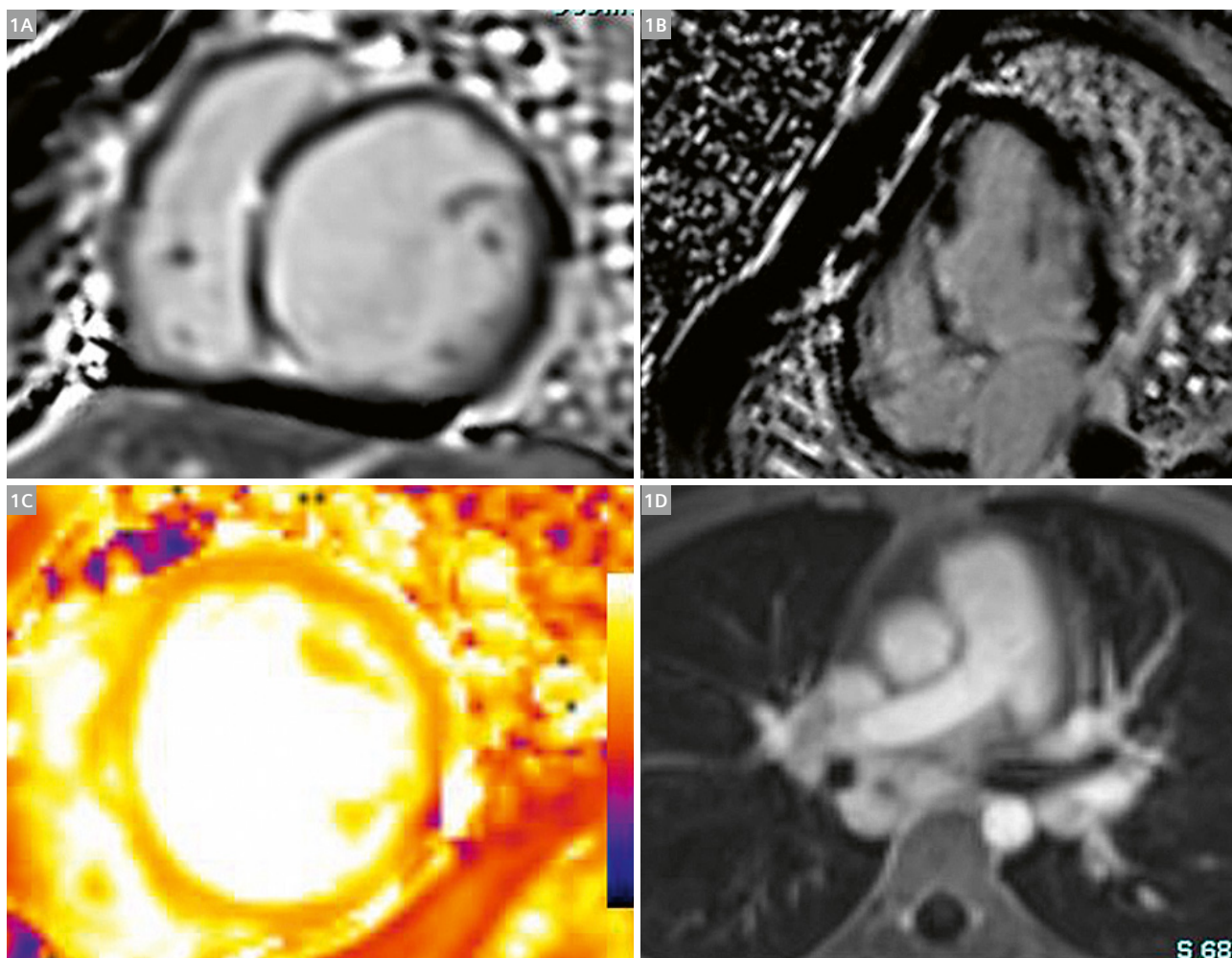
A 12-year-old female patient presented with dyspnea on exertion, giddiness, and palpitations. She had a history of loss of appetite and weight loss with low-grade fever and abdominal pain in the recent past. Multiple tests and hematological workup failed to reveal significant abnormalities.

She was admitted to the hospital with a syncopal episode. 2D echocardiography during this hospital admission revealed reduced ejection fraction and global hypokinesia. Her serum troponin levels were also elevated, so further workup was performed with cardiac MRI.

The results revealed global hypokinesia and severely reduced systolic function of the left ventricle. Multiple

small patchy foci of late gadolinium enhancement were seen in the septum (Fig. 1), and on the inferior and lateral walls. These were non-coronary territorial in distribution and showed elevated values on T1 and T2 mapping. Multiple conglomerated necrotic mediastinal lymph nodes were also demonstrated.

Further workup with tracheobronchial lymph node biopsy was performed. It revealed non-necrotizing epithelioid cell granulomas of mycobacterial etiology. The patient was started on antitubercular medications and heart failure management. The follow-up study showed significant clinical improvement and increase in ejection fraction.



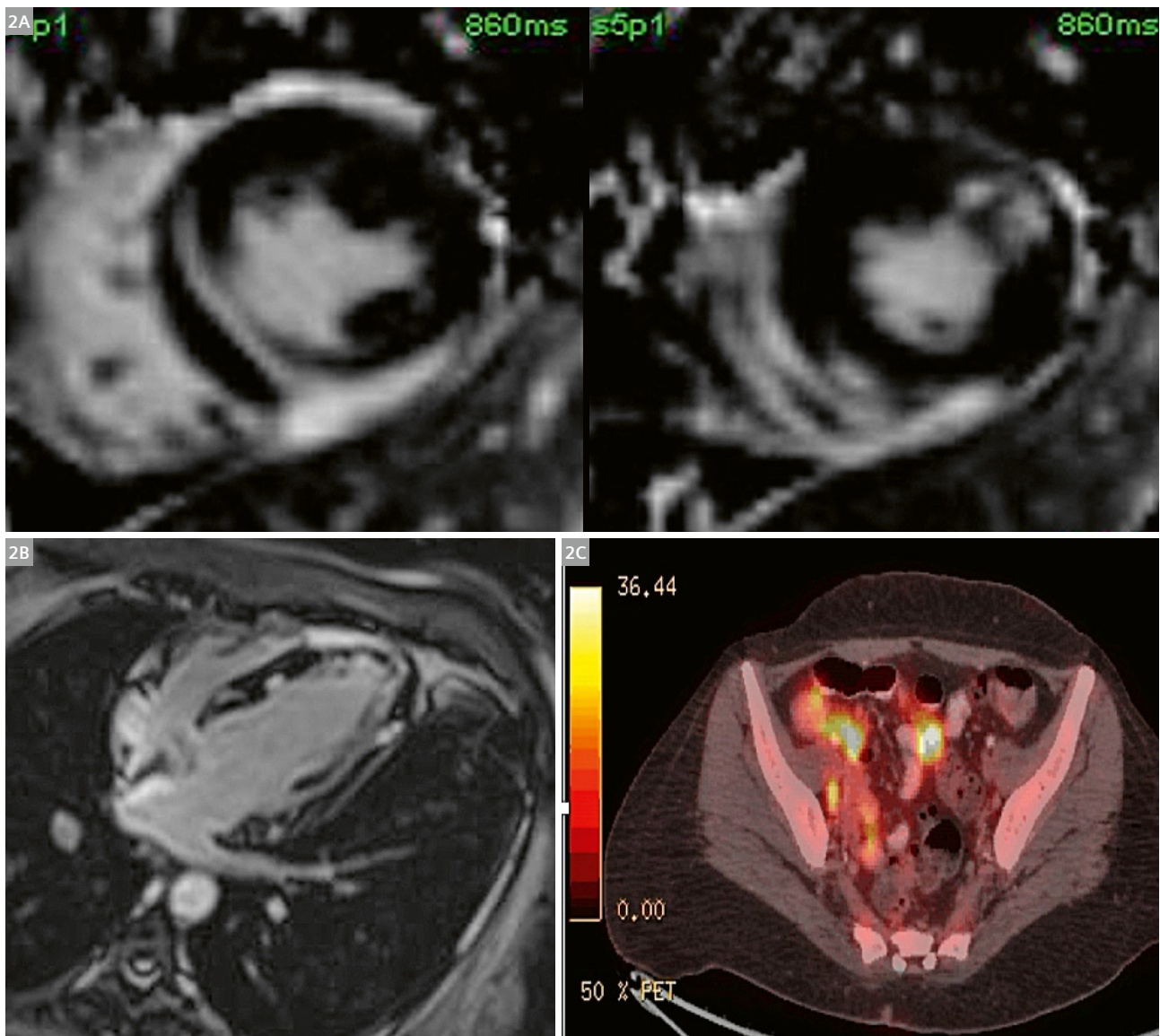
1 Patchy areas of late gadolinium enhancement are seen on phase-sensitive inversion recovery (PSIR) images (**1A** and **1B**) with elevated values on T1 mapping (**1C**). This non coronary territorial distribution and the necrotic mediastinal lymph nodes (**1D**) raised the possibility of tuberculosis, which was confirmed with lymph node biopsy.

Case 2

A 38-year-old female patient presented to the emergency department with cardiogenic shock. She was stabilized with DC cardioversion. Her serum troponins were significantly elevated, so cardiac MRI was performed for further workup. This revealed preserved function and extensive mid-myocardial and epicardial late gadolinium enhancement. This involved the anterior wall, septum, and the lateral and inferior walls. These segments revealed elevated values on T1 and T2 mapping. No mediastinal

lymph nodes were present. However, fibrotic changes were observed in the right upper lobe of the lung (Fig. 2).

After insertion of an implantable cardioverter defibrillator (ICD), a PET-CT exam was performed to evaluate the inflammatory cardiomyopathy. It showed multiple lymph nodes in the abdomen with high uptake in the right ileocecal region. A biopsy of these abdominal lymph nodes confirmed tuberculosis, and the patient was started on antitubercular therapy.

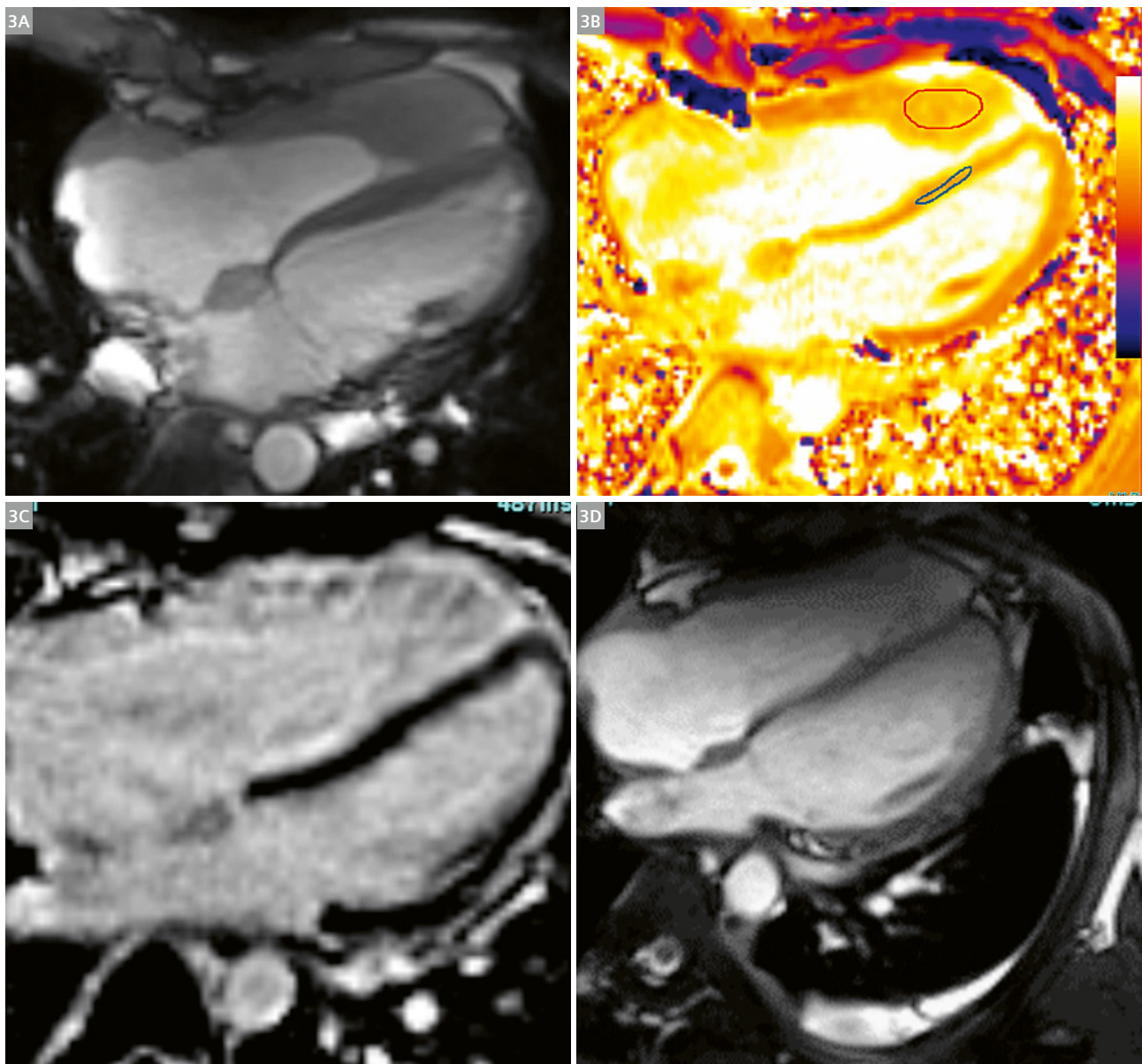


2 Multifocal patchy areas of mid-myocardial and epicardial late gadolinium enhancement were visualized on short-axis (2A) and 4-chamber-view (2B) images. PET imaging showing increased uptake in the right iliac fossa (2C) and abdominal lymph nodes. Biopsy confirmed tuberculous pathology.

Case 3

A 30-year-old male patient with massive tuberculous pleural effusion was evaluated using CT imaging. An incidental note of soft-tissue abnormality along the free wall of the right ventricle was made, so further workup with cardiac MRI was performed. This revealed soft-tissue thickening along the free wall of the right atrium and right ventricle with vascularity on perfusion images, and heterogeneous post-contrast enhancement. An incidental note of

focal osteolytic lesion in the dorsal vertebra was also made (Fig. 3). The patient was not willing to undergo an invasive procedure like endomyocardial biopsy. However, aggressive treatment with second-line antitubercular therapy was started in view of the tuberculous pleural effusion. A follow-up study revealed significant regression of the right ventricular abnormality. Clinical improvement was observed, along with an increase in ejection fraction.



3 Soft-tissue thickening along the free wall of the right atrium and ventricle (3A), which showed increased values on T1 mapping (3B), and late gadolinium enhancement on PSIR images (3C). A follow-up study after antitubercular treatment showed significant resolution (3D). Note the residual pleural effusion.

Comments

Myocardial involvement with tuberculosis is rare, and potentially treatable. Prompt recognition of this entity is crucial to prevent sudden cardiac death. High clinical suspicion and relevant investigations are necessary, along with cardiac MRI workup. Antitubercular therapy can result in successful regression of the disease and termination of malignant arrhythmia. More research is required to study this rare disease.

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Advanced Imaging in Cardio-Oncology: Cardiac Magnetic Resonance for Cardiotoxicity Assessment

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Cardiotoxicity: Background and definition

In recent decades, advancements in cancer therapies have significantly improved survival rates, leading to a growing number of cancer survivors. However, these successes highlight the need to address therapy-related side effects, particularly those affecting cardiovascular health, which can arise both acutely and chronically [1]. This recognition has fostered the development of cardio-oncology, a field dedicated to the cardiovascular implications of cancer treatments. Cardio-oncology employs integrated strategies for the prevention and management of associated risks, ultimately aiming to enhance cancer patients' quality of life [2].

The field adopts a multidisciplinary approach, with collaboration among various healthcare professionals to tackle the unique challenges faced by patients. Radiology plays a crucial role, providing essential diagnostic and monitoring tools that facilitate the early detection and characterization of cardiotoxicity. Advanced imaging modalities enable clinicians to identify subtle changes in cardiac function and structure, allowing for timely interventions that can prevent irreversible cardiac damage.

Historically, confusion has existed regarding the definitions of cardiotoxicity. This is due to the varied terminology used to describe the cardiovascular impacts of cancer treatments. Until 2022, the only widely accepted definition of treatment-induced cardiac dysfunction was from the American Society of Echocardiography, which defined it as a reduction in left ventricular ejection fraction (LVEF) of more than 10%, falling below 53% [3]. This definition has since broadened to encompass a wider array of cardiac issues stemming from cancer therapies.

In 2022, the International Cardio-Oncology Society (IC-OS) introduced a consensus definition for cancer therapy-related cardiac dysfunction (CTRCD), which

includes conditions such as cardiomyopathy, heart failure (HF), myocarditis, vascular toxicity, hypertension (HTN), arrhythmias, and QT interval prolongation associated with anticancer treatments [4]. This comprehensive framework aims to address long-standing issues of misclassification and misdiagnosis in the field. In August 2022, the European Society of Cardiology (ESC) published its first guidelines on cardio-oncology, establishing international standards for the diagnosis and management of cancer therapy-related cardiovascular toxicity (CTR-CVT). This initiative involved collaboration with the European Hematology Association (EHA), the European Society for Therapeutic Radiology and Oncology (ESTRO), and the International Cardio-Oncology Society, highlighting the need for a systematic approach to managing cardiovascular toxicities in cancer patients [5]. The guidelines stress the importance of early detection, risk assessment, and preventive measures to improve outcomes for this at-risk population.

Pathophysiology of cardiotoxicity

Cardiotoxicity arises from complex, multifactorial mechanisms that include direct myocardial injury, vascular endothelial damage, and inflammation, leading to structural and functional cardiac impairment [6].

Oxidative stress and mitochondrial dysfunction

Anthracyclines induce oxidative stress, generating reactive oxygen species (ROS) that damage cardiac myocytes. The heart's reliance on mitochondria makes it vulnerable. ROS accumulation leads to mitochondrial dysfunction, impaired adenosine triphosphate (ATP) synthesis, cardiomyocyte apoptosis, myocardial atrophy, fibrosis, contractile dysfunction, and heart failure [7].

Inflammatory pathways and immune activation

Immunotherapy, like checkpoint inhibitors (ICI), improves outcomes in malignancies but can cause immune-related adverse events (IrAEs), including cardiotoxicity in up to 1% of patients. Symptoms include hypotension, arrhythmias, and left ventricular dysfunction linked to cytokine release syndrome. Acute myocarditis occurs in 0.04% to 1.14% of cases, with risks ranging from fatal outcomes to transient edema within one to two months of treatment [8]. Mechanisms involve T cells targeting shared antigens in tumors and heart tissues, pericardial inflammation, myocardial ischemia affecting coronary plaques, and coronary spasms post PD-1 inhibitor treatment [9]. Imaging, especially cardiac magnetic resonance (CMR), aids in diagnosing cardiovascular adverse reactions as per the 2022 ESC guidelines, and offers advantages in identifying myocarditis.

Microvascular injury

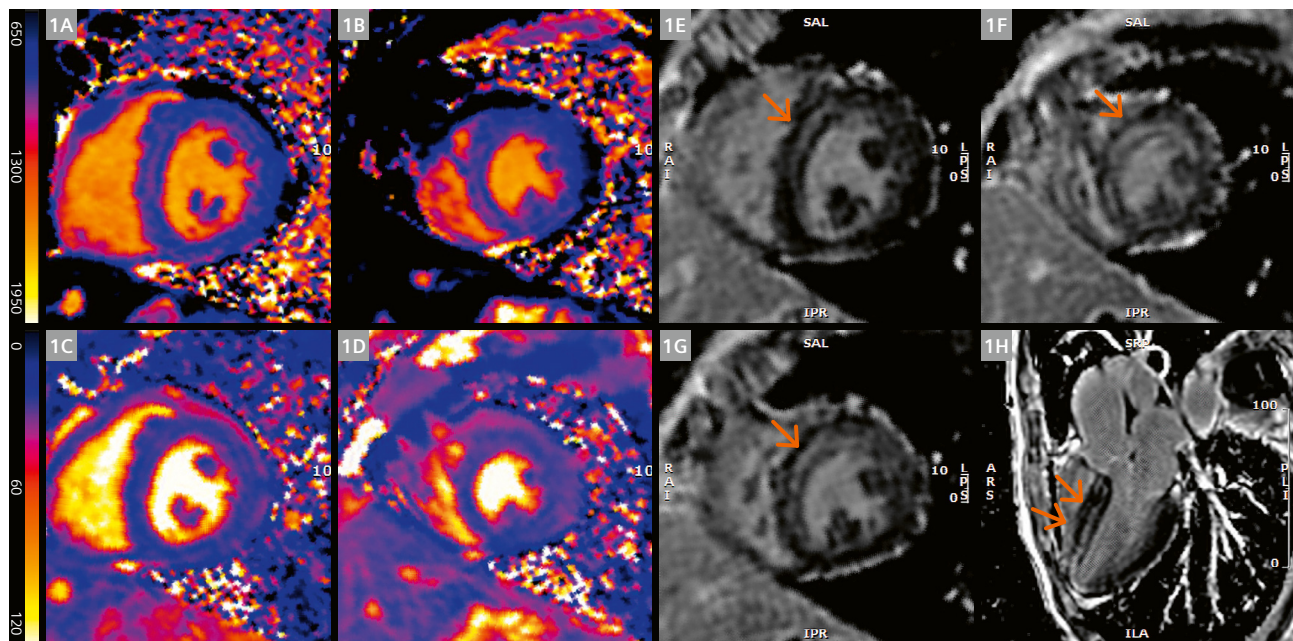
Tyrosine kinase inhibitors (TKIs) induce endothelial dysfunction and microvascular damage, reducing coronary flow reserve and causing vasospasm, increased vascular permeability, and myocardial ischemia [10].

The role of imaging in cardio-oncology

Advanced cardiovascular imaging plays a major role in detecting subclinical cardiovascular diseases and directly influences clinical decisions. The guidelines advocate for transthoracic echocardiography (TTE) as the primary imaging technique for baseline evaluations of heart structure and function before chemotherapy [11]. TTE quantitatively assesses left and right ventricular function, chamber dilation, hypertrophy, regional wall motion abnormalities, diastolic function, valvular heart disease, and pulmonary arterial pressure – all of which can guide therapeutic strategies.

The guidelines highlight the need for consistent imaging modalities throughout the treatment process to facilitate accurate comparisons over time. While TTE is crucial for assessing left ventricular function and global longitudinal strain (GLS) [12], CMR should be used when TTE yields poor quality images or non-diagnostic results [13].

In circumstances where TTE fails to identify specific conditions like hypertrophic cardiomyopathy, CMR becomes essential for thorough risk assessment. Both cardiac computed tomography angiography (CCTA) and CMR are indicated for detecting subclinical coronary artery disease (CAD) through coronary calcium scoring and identifying



1 A 65-year-old woman with Stage IV clear cell renal carcinoma developing immuno-checkpoint-inhibitor-related myocarditis. The CMR shows focally increased T1 values (1090 msec; normal value < 1059 msec; **1A, 1B**) and T2 values (55 msec; normal value < 51 msec; **1C, 1D**) at midventricular anteroseptal and apical septal wall. Late gadolinium enhancement (LGE) sequences show mid-myocardial wall LGE at the same level, as seen in the midventricular and apical short-axis view (orange arrows in **1E–1G**) and three-chamber view (orange arrows in **1H**).

intracardiac masses [14]. For symptomatic CAD, functional imaging techniques such as stress CMR are warranted, while CCTA serves as the standard test to exclude coronary obstructions in patients at low to intermediate risk of CAD [15].

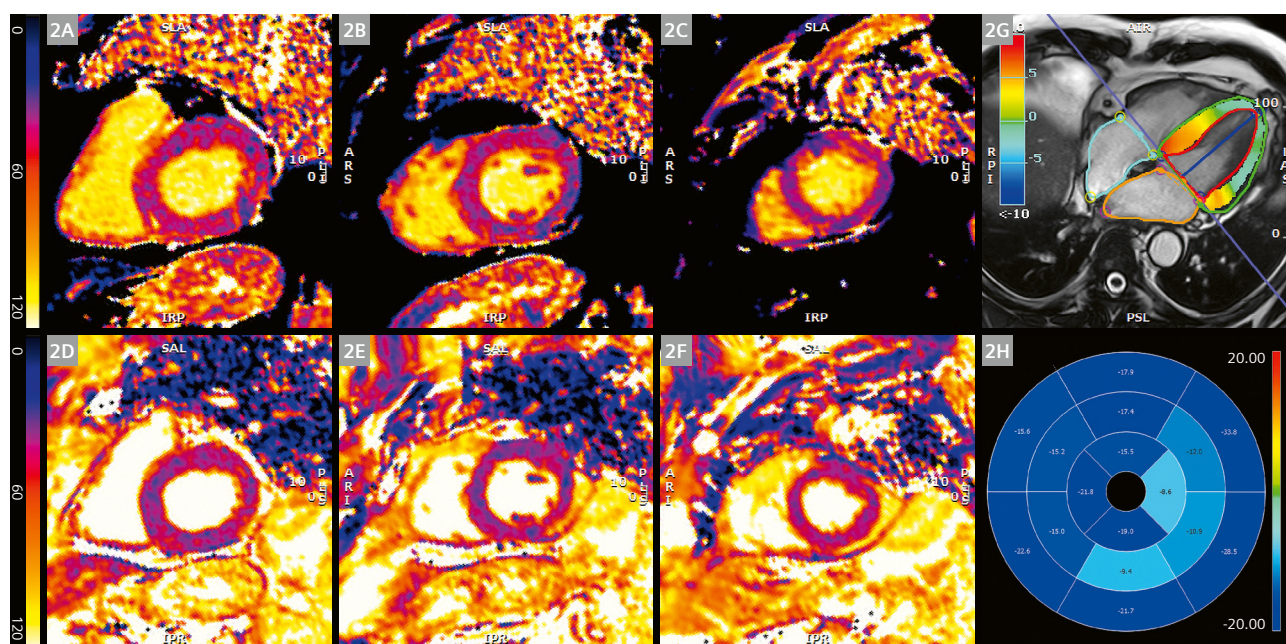
Role of CMR in tissue characterization and functional evaluation

Thanks to its spatial and temporal resolution, CMR is an invaluable tool for assessing cardiac structure and function without reliance on geometric assumptions – unlike other imaging modalities. It excels in characterizing myocardial tissue abnormalities stemming from chemotherapy, which can range from edema and necrosis to more chronic changes such as fibrosis. The use of advanced multiparametric CMR imaging allows clinicians to explore various aspects of cardiotoxicity through a combination of conventional and innovative T1/T2 mapping techniques [16–20]. Quantitative T1 and T2 mapping techniques in CMR have proven useful in reflecting subclinical histopathological changes in anthracycline-induced cardiotoxicity, aiding in early detection and management [21].

As well as providing critical insights into myocardial edema and inflammation, the mapping techniques also quantify extracellular volume (ECV), which serves as

an indirect biomarker for tissue fibrosis and interstitial expansion. This quantitative approach is vital for comprehensively evaluating the extent of myocardial damage, particularly in cases of diffuse myocardial disease, where regional changes may escape detection with standard imaging techniques like late gadolinium enhancement (LGE). Generally, an increase in pre-contrast T1 values indicates myocardial edema inflammation, and fibrosis, while prolonged T2 relaxation time correlates with acute edema. In several cases, cardiotoxic drug exposure has been associated with an increase in both native T1 and T2 relaxation times. Tissue characterization in cardiac imaging also employs qualitative and semi-quantitative methods that involve the use of gadolinium contrast agents. Early gadolinium enhancement (EGE) and late gadolinium enhancement are critical indicators of underlying pathological processes such as inflammation, edema, and myocardial fibrosis, all of which are associated with cardiotoxic chemotherapy [22]. Research has shown that EGE can predict a decline in left ventricular ejection fraction within the first month of anthracycline therapy. Additionally, the LGE pattern seen in trastuzumab- and anthracycline-induced myocardial fibrosis is characterized as non-ischemic subepicardial contrast enhancement [23].

Another significant factor in tissue assessment is extracellular volume, which effectively detects diffuse



2 A 45-year-old woman with breast cancer undergoing anthracycline chemotherapy. The CMR shows diffusely increased global T1 value (1045 msec; normal value < 1031 msec; **2A–2C**), with normal global T2 value (**2D–2F**) and mildly reduced global longitudinal strain (GLS = –17%; **2G** and **2H**).

myocardial fibrosis that may not be easily identified using LGE alone. Studies indicate a strong correlation between ECV measurements and histological specimens, highlighting its reliability [24]. In the acute phase, an increase in ECV may result from inflammation and interstitial edema [25]. Over time, persistently elevated ECV values are likely due to the progression of edema into interstitial fibrosis, providing crucial insights into myocardial health and the long-term effects of treatment.

Value of tracking CMR features

Subtle early tissue changes following chemotherapy can result in regional wall motion abnormalities, which can be detected using advanced cardiac magnetic resonance (CMR) strain imaging. Global longitudinal strain (GLS) is increasingly recognized as a more accurate measure of left ventricular dysfunction compared to traditional left ventricular ejection fraction assessments [26]. Several studies have shown that both GLS and global circumferential strain (GCS) are critical for identifying early signs of cardiotoxicity, particularly in patients treated with anthracyclines and trastuzumab, where reductions in GLS and GCS may occur alongside minor decreases in LVEF [27]. The assessment of deformation parameters like GLS is essential for risk stratification of cancer therapy-related cardiovascular toxicity prior to the initiation of cardiotoxic treatments. GLS and GCS are preferred over global radial strain (GRS), as they have fewer technical limitations [28]. In patients undergoing anthracycline therapy, declines in LVEF and strain values have been associated with atrophic remodeling of the myocardium resulting from oxidative damage. Significant decreases in left ventricular mass have been observed after anthracycline therapy, indicating an increased risk of cardiac events and heart failure symptoms, which may be more pronounced than the changes seen in LVEF alone [29, 30].

The financial, operational, and clinical benefits of CMR in detecting chemotherapy-induced cardiotoxicity

CMR has become an essential tool in the early detection and management of chemotherapy-induced cardiotoxicity, offering significant financial, operational, and clinical advantages. By providing precise, non-invasive, and highly accurate cardiac assessments, CMR enhances patient care while reducing healthcare costs and improving operational efficiency in oncology and cardiology settings.

The integration of CMR into routine monitoring protocols for patients undergoing potentially cardiotoxic chemotherapy can lead to substantial healthcare cost savings. By accurately identifying early signs of cardiac dysfunction, CMR enables timely interventions that may prevent progression to severe cardiotoxicity, thereby reducing the need for expensive treatments associated with advanced heart

failure [31]. Moreover, the precision of CMR decreases the reliance on multiple diagnostic tests, streamlining patient care and reducing expenses.

Operationally, CMR enhances diagnostic efficiency in the oncology setting. Its comprehensive imaging capabilities allow for the simultaneous assessment of cardiac structure, function, and tissue characterization in a single session, reducing the need for multiple imaging modalities. This efficiency optimizes workflow and resource utilization within healthcare facilities. Additionally, the non-invasive nature of CMR minimizes patient discomfort and risk, facilitating smoother operational processes.

Clinically, CMR offers unparalleled accuracy in detecting subclinical myocardial changes before the onset of overt cardiac dysfunction. Advanced techniques, such as myocardial strain analysis and tissue mapping, have demonstrated high sensitivity in identifying early myocardial injury. For instance, a cohort study highlighted the clinical usefulness of CMR-based tissue characterization for cardiotoxicity assessment, emphasizing its role in early detection and improved patient outcomes [32]. Early detection through CMR facilitates prompt cardioprotective interventions, potentially reversing myocardial damage and enhancing patient prognosis [33].

In summary, the integration of CMR into the diagnostic pathway for chemotherapy-induced cardiotoxicity offers comprehensive benefits. Its ability to provide early and accurate detection not only enhances patient care, but also contributes to the financial sustainability and operational efficiency of healthcare systems.

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Integrating SAMER Retrospective Motion Correction into 3D Deep Learning Image Reconstruction for High-Quality, Fast, and Robust Brain MRI

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Introduction

With the increasing use of medical imaging and with declining reimbursements, there is a growing demand for high-quality, fast, and robust MRI techniques. Patient motion is one of the most common sources of image degradation in clinical neuroimaging, manifesting as blurring, ringing, or in severe cases ghosting/folding artifacts. Radiologists are often forced to either interpret images despite these artifacts – risking missed or misinterpreted pathologies – or to request costly and time-consuming repeat scans. In pediatric patients, motion is typically more severe, often necessitating anesthesia, which increases both procedural risks and costs [1]. Additionally, motion artifacts can compromise the performance of automated quantitative clinical support tools [2], such as those used for brain morphometry and the identification or segmentation of hemorrhages, edema, and tumors. These tools are becoming increasingly critical in the screening, monitoring, and treatment of neurodegenerative diseases such as Alzheimer's.

Recent advances in deep learning-based image reconstruction have significantly accelerated scan times while preserving high image quality [3]. These improvements not only enhance scanning efficiency and patient throughput, but also reduce patient discomfort and the likelihood of motion-related artifacts. However, faster imaging alone is not enough to completely eliminate motion artifacts [4]. This challenge can be mitigated by combining fast imaging with motion-robust acquisition techniques such as BLADE/

PROPELLER [5], and/or directly applying motion correction strategies. The latter approach requires motion tracking data, which can be acquired through various methods, including optical tracking [6], navigator-based techniques [7], or data-driven approaches [8, 9]. These techniques can be applied prospectively – adjusting the field of view in real time during acquisition – or retrospectively during image reconstruction.

Scout accelerated motion estimation and reduction (SAMER) [10, 11] is a retrospective motion correction technique for brain imaging that enables fast motion estimation and artifact correction without the need for external tracking hardware. Clinically evaluated in both adult [12, 13] and pediatric [14] patient populations, SAMER has provided clinical value by significantly reducing the number of non-diagnostic motion cases. More recently, it has also been integrated into a deep learning-based image reconstruction framework to support highly accelerated, motion-robust 3D brain imaging [15]. In this article, we review the combined DL-SAMER¹ technique and demonstrate its effectiveness in vivo using MPAGE and SPACE acquisitions at R = 6 acceleration.

SAMER motion estimation

SAMER requires only small modifications to standard data acquisitions. In 3D multishot sequences, such as MPAGE or SPACE, a limited number of additional *k*-space encoding

¹DL-SAMER is work in progress. The application is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

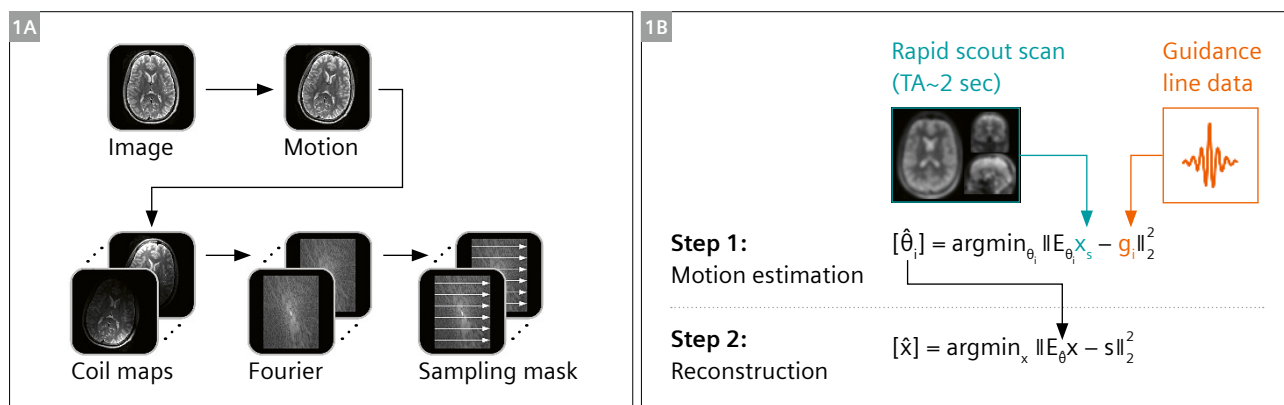
lines – referred to as motion guidance lines – are acquired within each repetition time (TR) of the data acquisition. Additionally, an ultra-fast, low-resolution scout scan (~2 seconds), designed to match the contrast of the guidance lines, is performed once before each imaging sequence. Rapid motion trajectory estimation is achieved by minimizing the discrepancy between a rigid-body physics-based forward model (Fig. 1A) – which leverages the scout scan as an image prior – and the acquired guidance line k -space data (Fig. 1B).

The key computational advantage of SAMER is its ability to estimate motion for each TR independently, without requiring intermediate image volume reconstructions. This enables extremely fast (GPU) motion estimation (~1 second per shot) using standard scanner graphics processing unit hardware. Moreover, the per-shot motion optimizations can be initiated during the acquisition. This makes efficient use of the scanning time and allows the

final motion-corrected image reconstruction to start shortly after scan completion.

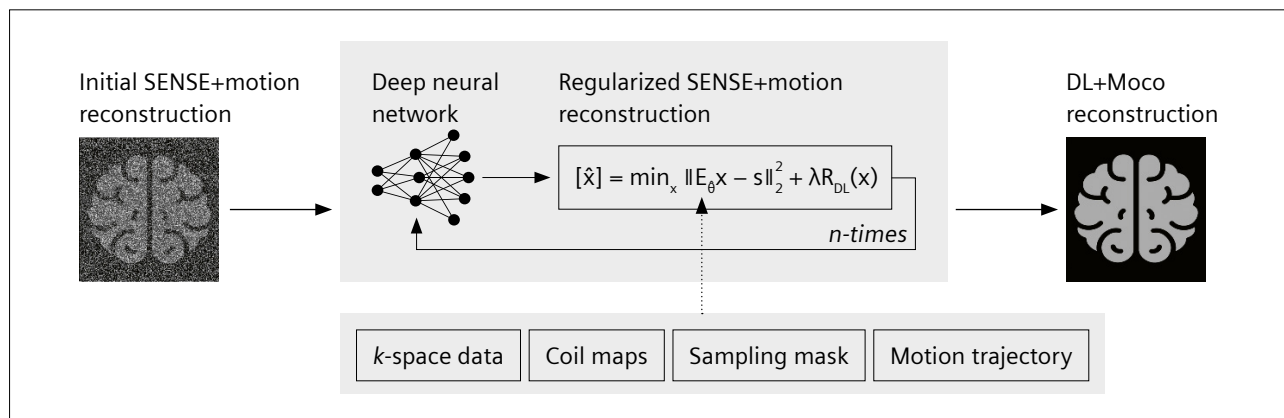
Deep learning image reconstruction

Deep learning was integrated into the SAMER motion-mitigated image reconstruction framework (Fig. 2) to enable highly accelerated, motion-robust 3D imaging. This approach leverages a state-of-the-art unrolled network architecture, which alternates between classical conjugate-gradient SENSE+motion optimization and a deep learning-based update of the image prior [3]. At each iteration of the reconstruction, a deep neural network is used to suppress noise and residual artifacts from the current image estimate. The network output then serves as a prior in a regularized SENSE+motion reconstruction, and the final image is obtained after a fixed number of reconstruction steps.



1 Illustration of SAMER retrospective motion correction.

(1A) Physics-based SENSE+motion model E_{θ} describes the effect of motion on acquired k -space data. (1B) SAMER scout x_s and acquisition of additional k -space encoding lines g_i (motion guidance lines) enable very rapid and fully separable estimation of motion parameters shot-by-shot (Step 1). Estimated trajectory θ is used to reconstruct the motion-corrected image (Step 2).



2 Illustration of unrolled deep learning reconstruction with integrated motion correction.

A deep neural network is used to regularize a motion-informed conjugate-gradient reconstruction (SENSE+motion). The final DL+motion correction (Moco) image is obtained after n -iterations of the network-regularized reconstruction.

Applications

The SAMER framework was integrated into 3D MPRAGE and 3D SPACE FLAIR sequences. Scans were conducted on a healthy volunteer using a 3T MRI scanner (MAGNETOM Vida; Siemens Healthineers, Erlangen, Germany) with a 20-channel head coil. The acquisition parameters were as follows:

- MPRAGE: $1 \times 1 \times 1 \text{ mm}^3$, TR 2300 ms, TI 900 ms, TE 2.22 ms, iPAT 3×2 (CAIPI), TA 1:53 min
- SPACE FLAIR: $1 \times 1 \times 1 \text{ mm}^3$, TR 5000 ms, TI 1800 ms, TE 389 ms, iPAT 3×2 (CAIPI), TA 1:47 min

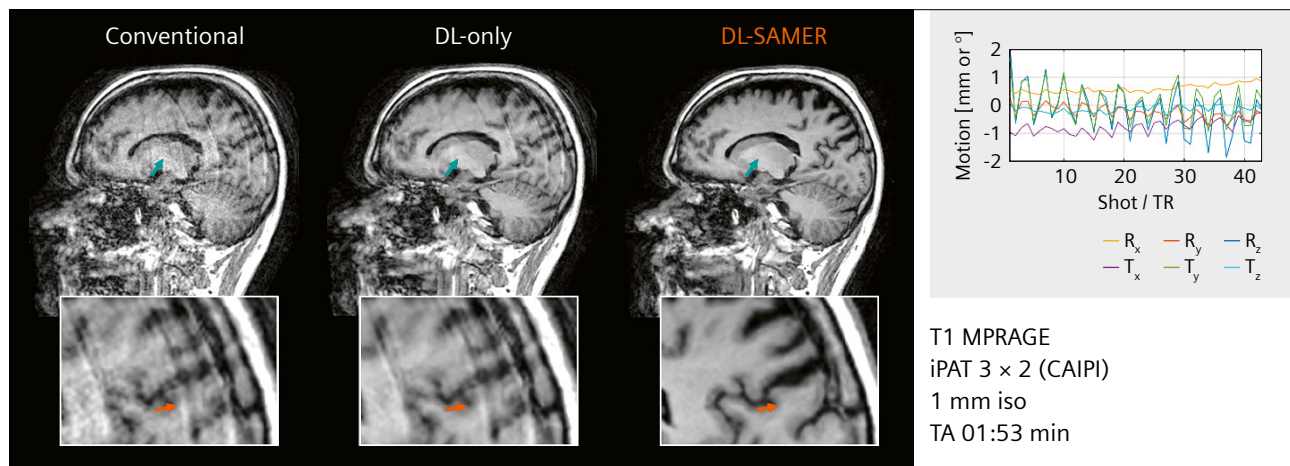
The volunteer was trained to perform motion patterns resembling clinical scenarios, including periodic nodding to

simulate heavy breathing and free, unsupervised motion.

All scans were reconstructed using the following:

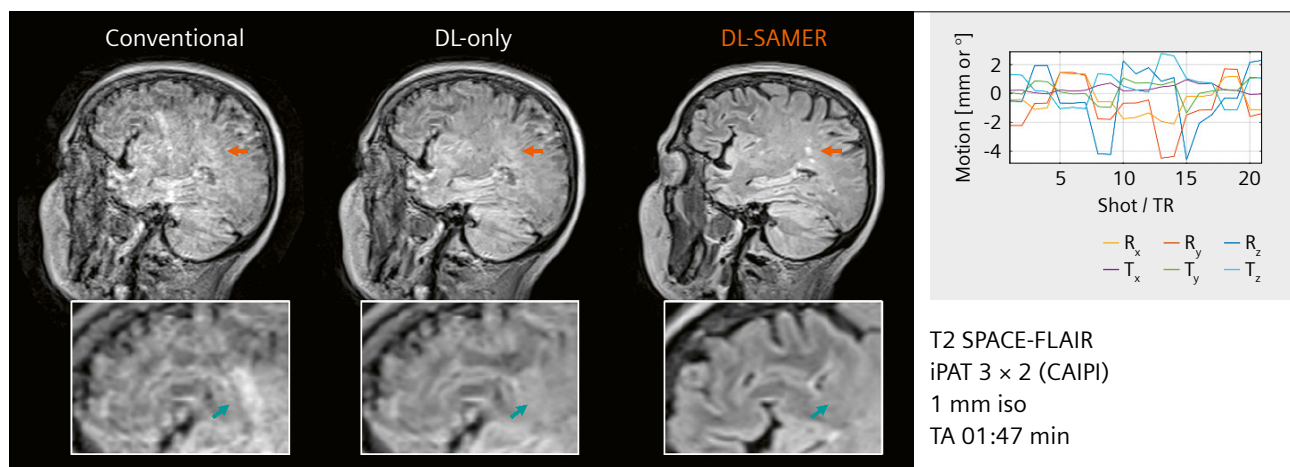
- 1) A SENSE-based reconstruction ("conventional")
- 2) A deep learning-based reconstruction ("DL-only")
- 3) A combined deep learning and motion-corrected reconstruction ("DL-SAMER")

Figure 3 shows reconstructions of an $R = 3 \times 2$ accelerated 3D MPRAGE acquisition with instructed heavy nodding to simulate breathing motion. The DL-only method effectively reduced parallel imaging artifacts and improved the signal-to-noise ratio (SNR) compared to the conventional SENSE reconstruction. However, ringing artifacts caused by sub-



3 Reconstructions of $R = 3 \times 2$ accelerated 3D MPRAGE with instructed heavy nodding (breathing) motion.

DL-only improved SNR (petrol arrows), but motion artifacts remain (orange arrows). DL-SAMER improved SNR and increased image sharpness through a motion-informed 3D deep learning image reconstruction.



3 Reconstructions of $R = 3 \times 2$ accelerated 3D SPACE FLAIR with free, unsupervised motion.

The combined DL-SAMER approach reduced aliasing artifacts (petrol arrows) and improved image sharpness, as observed by better depiction of white matter hyperintensities (orange arrows).

ject motion remained. In contrast, the combined DL-SAMER approach successfully mitigated most motion artifacts, resulting in enhanced image sharpness while maintaining a high SNR.

Figure 4 contains reconstructions of $R = 3 \times 2$ accelerated 3D SPACE FLAIR data acquired during free, unsupervised motion. The DL-SAMER approach effectively reduced both aliasing and motion artifacts, leading to improved image sharpness. This is evident in the better delineation of gray-white matter boundaries and the clearer depiction of white matter hyperintensities, which are not visible in the DL-only reconstruction.

Conclusions

DL-SAMER combines deep learning-based image reconstruction with SAMER motion correction, enabling fast and motion-robust 3D brain imaging. The method generalizes across important clinical contrasts, including T1, T2, and T2-FLAIR, thus closely aligning with routine brain exams. By integrating deep learning into the reconstruction, DL-SAMER achieves high-quality, motion-corrected imaging with high SNR and significantly reduced scan times, and is expected to improve the reliability of automated quantitative clinical support tools when applied to challenging patient populations.

A clinical evaluation of DL-SAMER is currently underway in the inpatient and emergency settings at the Neuroradiology Department of Massachusetts General Hospital (Boston, MA, USA). Since SAMER's motion estimation operates independently of the deep learning-based motion-correction reconstruction, similar diagnostic quality improvements observed in previous SAMER studies [12, 13] – without deep learning and lower acceleration – are anticipated.

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Unlocking New Frontiers in Neurological Imaging: The Power of Second-Generation Clinical 7T MRI

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Abstract

With optimized imaging protocols, the enhanced signal-to-noise ratio and spatial resolution of second-generation 7T MRI enable superior detection of cortical lesions, microbleeds, and epileptic foci. These improvements in imaging may provide new avenues for targeted interventions and therapeutic decisions in neurological disorders, highlighting the potential of 7T MRI as a valuable tool in high-precision medicine.

Introduction

There are now more than one hundred 7-Tesla (7T) MRI scanners worldwide, with the majority installed in research settings. However, an increasing number of centers are using 7T MRI for routine clinical examinations [1]. From the perspective of conventional radiological imaging, the benefits of ultra-high-field imaging include the potential for higher spatial resolution due to improved signal-to-noise ratio, and enhanced tissue contrast arising from



1 Setup of the 7T MAGNETOM Terra.X for clinical neuroimaging. Patients were positioned head-first in the 8Tx/32Rx coil, which has parallel transmit capabilities.

increased susceptibility differences in tissues under the influence of a strong static magnetic field [2].

Numerous studies using the first generation of clinical 7T MRI scanners have demonstrated the advantages of high-field imaging. For instance, in patients with multiple sclerosis, 7T MRI reveals significantly more cortical and juxtacortical lesions than 3T MRI, helping physicians differentiate this demyelinating disease from other conditions, such as neuromyelitis optica spectrum disorder, which does not typically involve the cortex [3, 4]. The central venous sign, an imaging feature possibly relevant to the upcoming revised McDonald diagnostic criteria for multiple sclerosis, is also more clearly visualized at 7T MRI [5]. Furthermore, high-field MRI is particularly valuable in epilepsy for detecting small focal cortical malformations, hippocampal abnormalities, and encephaloceles [6].

Cerebral amyloid angiopathy (CAA), a subtype of cerebral small vessel disease, involves pathologically altered vessels smaller than 500 micrometers in diameter. It is caused by excessive beta-amyloid deposition, especially in the cortex and leptomeningeal vessels, and leads to microvascular disruption [7]. In these cases, 7T MRI offers superior identification, characterization, and detection of cerebral microbleeds, as well as potential associations with vascular abnormalities [8]. It also enhances the visualization and quantification of perivascular spaces and the detection of subarachnoid FLAIR hyperintensities, which may indicate vascular leptomeningeal leakage of plasma proteins or erythrocytes, serving as an early marker of developing superficial siderosis [9, 10]. In non-contrast arterial imaging, such as time-of-flight angiography, 7T MRI has also been shown to be superior in distinguishing true aneurysms from infundibular vascular outflows [11],

with resolutions comparable to those of digital subtraction angiography.

Increasingly, 7T MRI is being integrated into routine clinical neuroradiological imaging for the abovementioned conditions, and it is anticipated that specialized centers will increasingly use this technology for targeted neurological assessments as part of high-precision medicine. The advantages of high-field imaging are further exemplified in the MAGNETOM Terra.X scanner from Siemens Healthineers (Erlangen, Germany), the company's second-generation 7T system, which received FDA approval in March 2024.

In this article, we show the technical setup of a MAGNETOM Terra.X in Geneva, and present a few cases from routine clinical use.

Materials and methods

Four patients provided informed consent for MRI examinations at 3T and 7T. At 7T, we employed an 8Tx/32Rx coil array with parallel transmit capabilities for our neuroradiological examinations (NovaMedical, Wilmington, MA, USA; CE-certified in 2023). The setup was analogous to clinical 3T MRI neuroimaging examinations: Patients were positioned head-first in the head coil (Fig. 1). As part of the protocol optimization for the conditions mentioned above, which involved clinical measurement sessions of approximately one hour, we selected sequences that we believe offer an optimal balance between efficiency and image quality. Detailed information on the sequences is provided in Table 1. Both the 3T and 7T protocols were optimized to align with clinical standards and workflow requirements regarding time and spatial resolution.

Sequence/ orientation	Acceleration method, acquisition factor	TE (ms)	TR (ms)	BW (Hz/Px)	Slice thickness (mm)	Voxel size (recon in-plane, mm × mm)	Acquisition time (min:s)
T1 MP2RAGE/sag	CS 5	3.08	6000	240	0.66	0.6 × 0.6	7:34
T2 FLAIR/sag	CAIPIRINHA 6	264	8000	751	0.7	0.7 × 0.7	8:18
T2 TSE/tra	GRAPPA 4	62	8440	407	2.0	0.22 × 0.22	3:58
T2 TSE/cor	GRAPPA 4	62	8440	407	2.0	0.22 × 0.22	3:58
DWI RESOLVE/tra, b = 0 or 1000 s/mm ²	SMS 3	TE1: 66 TE2: 106	5450	1202	2.0	0.7 × 0.7	3:21
SWI/tra	GRAPPA 3	7	25	400	1.6	0.15 × 0.15	6:34
TOF/tra	CS 7.2	3.29	110	496	0.4	0.3 × 0.3	7:37

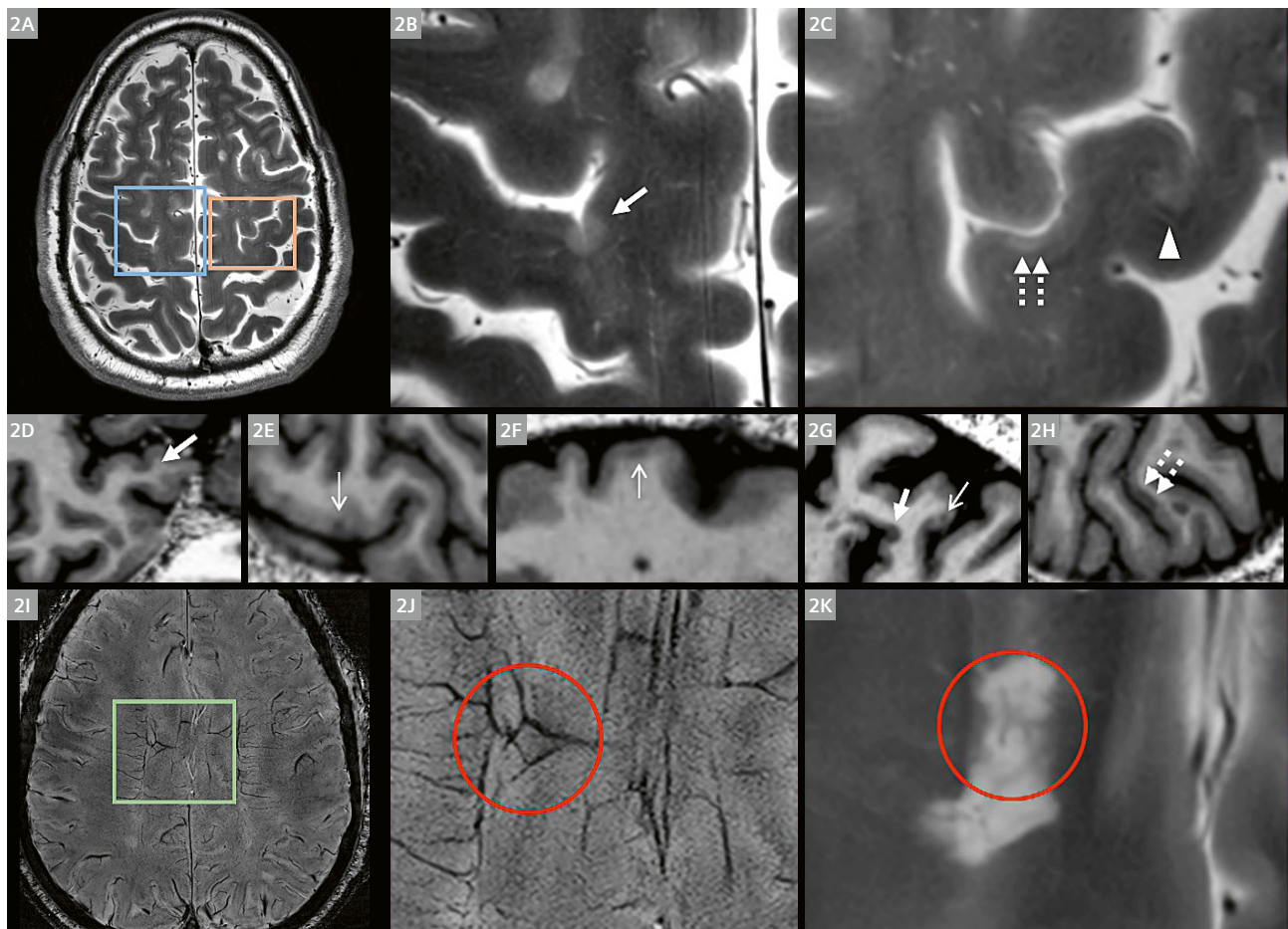
Table 1: Clinical 7T protocol sequence parameters.

Results

The images obtained with 7T MRI illustrate the benefits of second-generation high-field imaging for multiple sclerosis (Fig. 2), cerebral amyloid angiopathy (Fig. 3), epilepsy focus detection (Fig. 4), and aneurysm characterization (Fig. 5).

In our clinical protocol, the differentiation of juxtacortical (Type 1), purely cortical (Type 2), and subpial lesions (Type 3) in multiple sclerosis approaches the level of mesoscopic pathological examinations (Figs. 2A–2H) [12]. Subpial lesions, which extend from the superficial pial surface to deeper cortical layers, correlate most closely with the visualization of meningeal immune cell

aggregates – which are rich in B cells, plasma cells, and T cells – supported by follicular dendritic cells, indicative of focal meningeal inflammation [13]. In addition, 7T can effectively visualize and characterize venous abnormalities within multiple sclerosis lesions, which are potentially 'related to endothelial dysfunction due to a disease-related early increase in proinflammatory cytokines that leads to vascular remodeling, local hypoperfusion, and vessel wall breakdown with ensuing inflammatory changes in adjacent tissue [14] (Figs. 2I–2K).

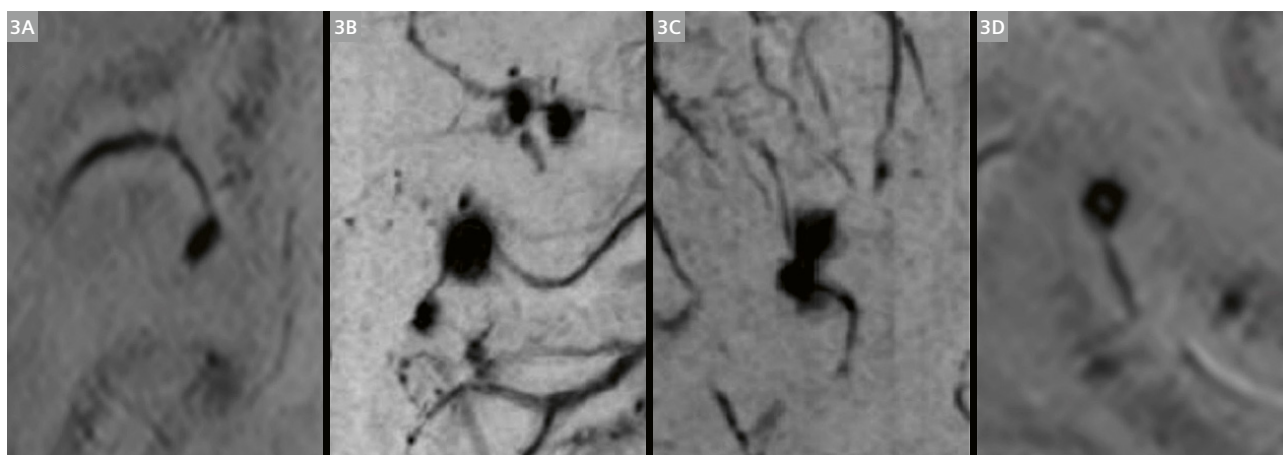


2 Cortical lesions and venous abnormalities in multiple sclerosis. (2A) T2-weighted image ($0.2 \times 0.2 \text{ mm}^2$ in-plane resolution) of the fronto-parietal brain parenchyma in a patient with multiple sclerosis. (2B) Enlarged blue-rectangle region from (2A) showing a cortico-juxtacortical lesion (white arrow), corresponding to a Type 1 cortical lesion. (2C) Enlarged orange-rectangle region from (2A) showing a subpial lesion (dashed arrows; Type 3 cortical lesion) and an irregularly shaped juxtacortical lesion. (2D–2H) T1-weighted images showing different cortical lesions, with thick white arrows indicating cortico-juxtacortical lesions (Type 1), thin white arrows indicating purely cortical lesions (Type 2), and dashed arrows indicating a subpial lesion (Type 2). (2I) SWI image ($0.15 \times 0.15 \text{ mm}^2$ in-plane resolution) of the same patient. (2J) Enlarged green-rectangle region from (2I) indicating a venous abnormality (red circle), possibly resulting from cytokine-induced vascular remodeling and associated subsequent local inflammatory parenchyma changes. (2K) Enlarged green-rectangle region from (2J) in the T2-weighted corresponding image, indicating a complex central vein sign.

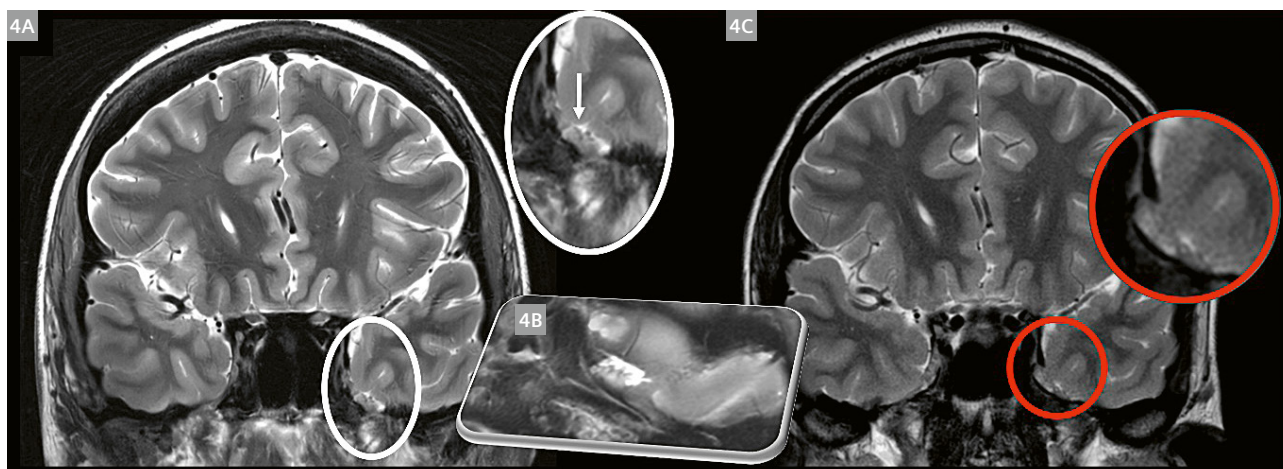
The increased signal-to-noise ratio at 7T MRI allows in-plane resolutions beyond $0.35 \times 0.35 \text{ mm}^2$, which recently enabled the first in vivo description of venous involvement in some cerebral microbleeds (CMB) in a small cohort of eight patients with cerebral amyloid angiopathy (CAA) [8]. In our study, we observed an association between small vessels and more than half of all microbleeds in patients with probable CAA according to the Boston Criteria v2.0 – compared to less than 20% reported in the literature – using an in-plane resolution of $0.15 \times 0.15 \text{ mm}^2$ (Fig. 3) [8].

Epilepsy focus detection can be a prolonged and often radiologically inconclusive process. Figure 4 presents the case of a patient with suspected left temporal lobe epilepsy based on EEG findings, but without visible focus on 3T MRI. Using our 7T MRI protocol, we identified a temporal encephalocele as the likely cause of the seizures. Patients with temporal encephaloceles generally have a favorable prognosis following targeted surgical intervention [15].

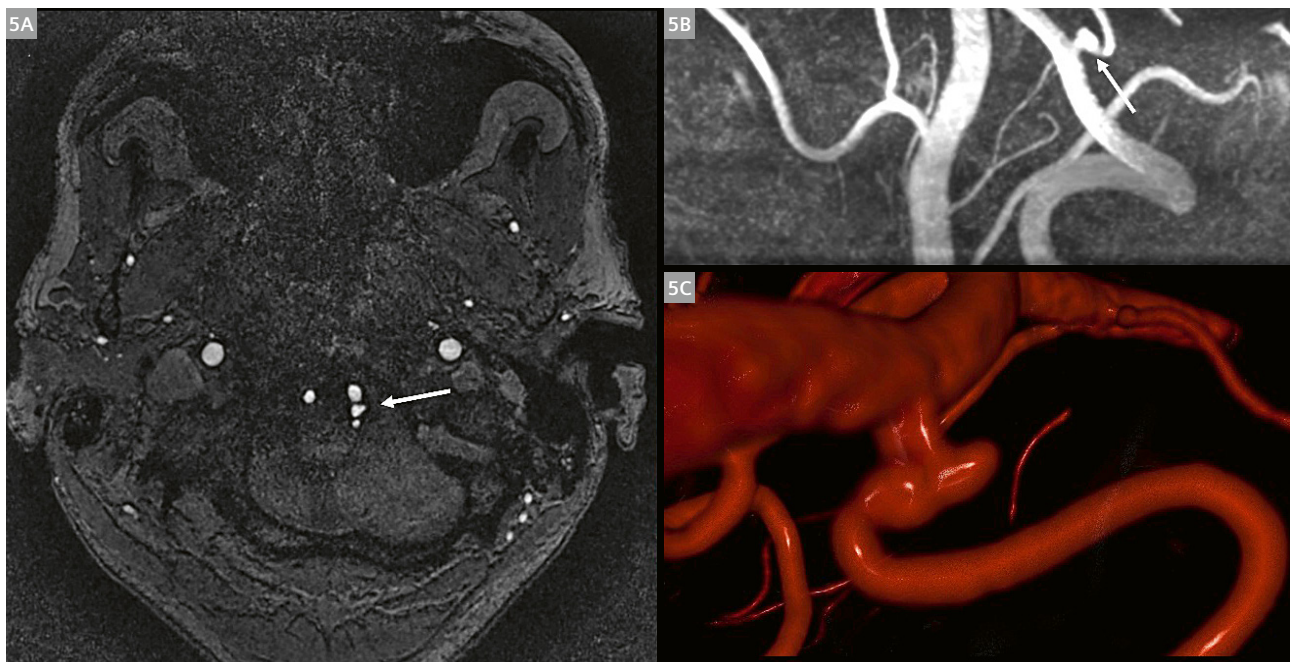
In Figure 5, 7T images of a lobulated vertebral aneurysm are shown, which was initially interpreted as an infundibular origin of the left posterior inferior cerebellar



3 Cerebral subcortical microbleeds with venous association in probable cerebral amyloid angiopathy (CAA). (3A–3D) SWI images ($0.15 \times 0.15 \text{ mm}^2$ in-plane resolution) of cerebral microbleeds in a patient with probable CAA according to the Boston Criteria v2.0. The second-generation 7T MRI allows improved hemorrhage detection and quantification, and improved analysis of microvascular abnormalities associated with cerebral microbleeds, potentially aiding in the study of CAA microbleed pathogenesis.



4 Encephalocele in temporal lobe epilepsy. (4A) T2-weighted coronal image ($0.2 \times 0.2 \text{ mm}^2$ in-plane resolution) in a patient with therapy-resistant epilepsy, with electroencephalogram studies indicating a left temporal lobe origin. A tiny encephalocele (white circle, white arrow) was found and confirmed on transverse T2-weighted imaging (4B). The same lesion was missed on preceding 3T MRI (4C; $0.4 \times 0.4 \text{ mm}^2$ in-plane resolution; red circle).



5 Aneurysm versus infundibulum of the left posterior inferior cerebellar artery (PICA). **(5A)** Time-of-flight image ($0.3 \times 0.3 \times 0.3 \text{ mm}^3$ resolution) indicating a vascular ectasia at the origin of the left PICA (white arrow) and **(5B)** maximum-intensity projection in a sagittal reconstruction showing the proximal PICA ectasia. **(5C)** 3D reconstruction indicates a lobulated aneurysm at the proximal left PICA.

artery on the 3T scan. This could pave the way for new, non-invasive diagnostic options in vascular surgery and interventional planning, potentially reducing the need for potentially harmful X-ray and contrast-agent-based procedures such as digital subtraction angiography.

Discussion and conclusion

The clinical use of 7T MRI remains limited to a few specialized centers, but the latest generation of 7T MRI scanners offers new diagnostic advantages that enable high-precision radiological imaging as part of routine clinical protocols. This has tangible therapeutic implications, such as the identification of additional demyelinating lesions on 7T MRI in multiple sclerosis patients, which could fulfill the McDonald criteria for spatial dissemination and guide therapeutic decisions, or the detection of epileptic foci leading to surgical intervention. Advances in parallel transmit applications and artificial intelligence-based techniques are expected to further enhance the efficiency of current protocols, shortening scan times and broadening access to these high-precision diagnostic tools in clinical practice.

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The CEA-Siemens Healthineers MAGNETOM Iseult 11.7-Tesla MRI System: A 20-year Odyssey to Image the Human Brain

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Ultra-high-field MRI: A tool to study the human brain

Understanding the human brain and its complex functions is instrumental for advancing medical science. Despite significant progress in neuroimaging [1, 2], many brain processes remain poorly understood due to their complexity and the limitations of current imaging technologies. Ultra-high-field (UHF) MRI offers supra-linear gains [3] in signal-to-noise ratio, which can be invested to achieve unprecedented spatial resolution [4, 5]. This allows for the visualization of smaller anatomical features, such as cortical layers and subcortical nuclei, which are often blurred in conventional MRI scans. By uncovering these features and processes, UHF MRI could contribute to a better global understanding of the physiopathology of several neurological disorders, and could help identify related biomarkers, leading to more accurate diagnoses and targeted treatments [6–8]. Reaching the “mesoscopic” scale in imaging [9] can help bridge the gap between molecular processes at the cellular level and whole-brain activity, fostering a more comprehensive understanding of the human brain.

Additionally, blood oxygenation level-dependent (BOLD) imaging [10] is based on variations in blood oxygenation, which are influenced by neuronal activity

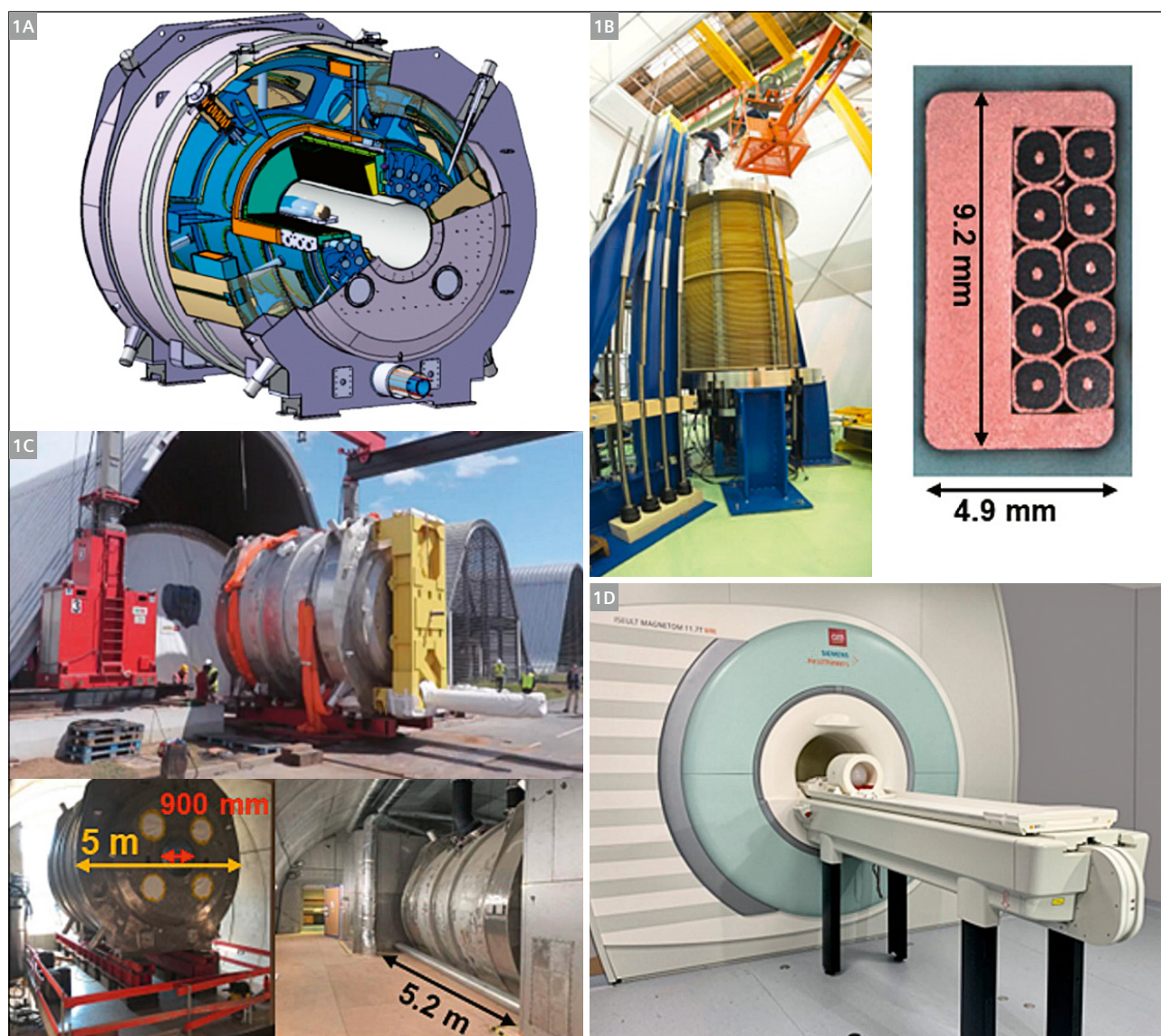
and are more detectable at UHF strengths. This boost in sensitivity, combined with increased spatio-temporal resolution, means BOLD significantly benefits from UHF MRI, enabling precise mapping of brain activity and providing insights into functional connectivity and neural pathways [11]. These advancements pave the way for studying more complex brain functions. Altogether, integrating UHF MRI into neuroscience research has the potential to revolutionize our knowledge of the brain and improve clinical outcomes for a variety of neurological conditions.

The Iseult project

Based on this idea, the Iseult project was launched in 2001 by the French Atomic Energy Commission (CEA) as an exceptional, visionary challenge. The project aimed to build an MRI magnet operating at 11.7T to investigate the human brain at a mesoscopic scale. After nearly 20 years of research and development mainly led by the Institute of Research into the Fundamental Laws of the Universe (IRFU) of the CEA, the 132-ton 11.7T magnet, with its imaging equipment, delivered its first images in 2021 [12].

The field homogeneity and temporal stability constraints for MRI were met thanks to a careful design of the magnet [13]. The core part of the Iseult MRI system is an actively shielded niobium-titanium (NbTi) magnet cooled with a superfluid helium (He) cryostable bath at 1.8 kelvin (K) [14], providing a homogeneous magnetic field within a 90 cm warm bore, operated in driven mode. A multi-strand wire magnet design was chosen to reduce the inductance of the main superconducting coil and thus reduce the maximum voltage induced in case of quench. The magnet

consists of 170 double pancake wires, with about 250 superconductive joints, making it a unique design. A magnetic field intensity of 11.7T was the upper limit to use an NbTi alloy at 1.8 K. A dedicated cryogenic plant was installed in the basement of the NeuroSpin building to service superfluid helium and cool down the magnet at 1.8 K. Figure 1 displays the magnet and associated devices, while detailed specifications of the magnet are provided in Table 1.



1 The Iseult magnet and installation set-up. **(1A)** Diagram of the magnet showing the main coil and the two actively shielding coils. The magnet weighs 132 tons, has an outer diameter and total length of just under 5 m, and the bore diameter is 90 cm. **(1B)** A view of the 170 double pancake wires at the manufacturing site in Belfort, France, and a cross-section of the magnet superconductor made of ten NbTi-Cu wires soldered into a copper channel. **(1C)** Installing the magnet in a dedicated NeuroSpin arch (10 m in diameter), and general views of the magnet inside the NeuroSpin arch. **(1D)** View of the front side of MAGNETOM Iseult¹, as seen by volunteers. A ramp is required to access the static patient table (not shown here).

¹MAGNETOM Iseult is an investigational device only and is not intended for sale or clinical use. Research operations may require observation of national regulations.

Current	1470 A
Operating temperature	1.8 K
Central field	11.72 T
Inductance	308 H
Stored energy	338 MJ
Mass	132 tons

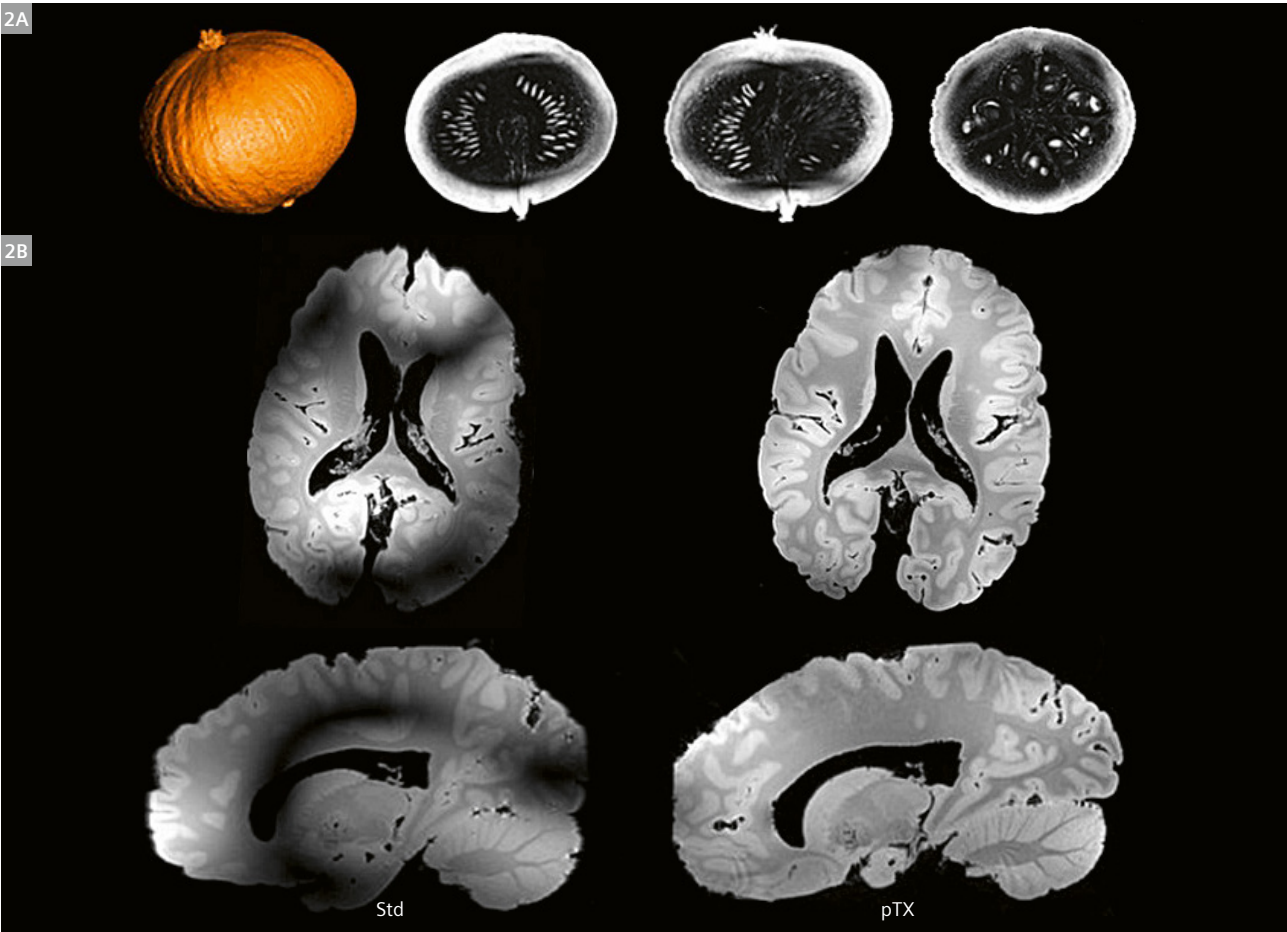
Table 1: Main parameters of the Iseult magnet.

The Iseult MRI system is currently equipped with a whole-body SC72 gradient coil (maximum gradient strength and slew rate: 80 mT/m and 200 T/m/s respectively; weight: 900 kg, length: 1.59 m, inner/outer diameter: 64/81 cm) designed and manufactured by Siemens Healthineers. A gradient coil-magnet interaction test campaign was carried out over a two-year period [12]. Gradient coils operate

under oscillating currents, which generate eddy currents in the different metallic shells under a strong magnetic field, inducing significant forces and vibrations. If not well characterized and understood, their consequences can be disastrous and lead to magnet quench, hardware damage, or image artifacts. The objective of the test campaign was therefore:

- 1) to determine safe and optimal MR exploitation modes in terms of gradient frequency, strength, and duration;
- 2) to gather information to troubleshoot the system if artifacts are encountered during imaging. The tests covered acoustics, vibrations, power deposition in the helium bath, and field monitoring [15] measurements. For a comprehensive description and results of this study, see [12].

The last piece of equipment is a geometrically decoupled 16Tx/32Rx radiofrequency (RF) custom coil (combined to 8Tx/32Rx), which the CEA designed specifically for the Iseult 11.7T project [16]. The transmit-receive array is made of 15 linear resonators (dipoles) arranged in lower/



2 (2A) Gradient-echo (GRE) imaging of a pumpkin with a 400 μm isotropic resolution, acquired with the service coil from Siemens Healthineers (QED, birdcage architecture). The acquisition duration was about 4 hours (no acceleration). (2B) 3D GRE images acquired with the Iseult parallel transmission (pTx) RF coil on an ex vivo brain, with phase-shimming (left) and kT points (right). Axial (top) and sagittal (bottom) slices are shown, demonstrating the significant improvement in image quality with pTx.

upper rows in z-direction, and one patch resonator, while the receive array consists of 16 loops. Eight RF power amplifiers feed the RF transmit elements to mitigate the RF field inhomogeneity problem. This peculiar architecture satisfies two needs: compactness that enables compatibility with a local B0 shim insert [17], and easy tuning and performance prediction. Figure 2 shows the first images acquired with the system at the end of 2021.

Patient-Reported Experience Measures

Before the start-up of MAGNETOM Iseult¹, the highest magnetic field used for MRI on humans was 10.5T, at the University of Minnesota in the United States [18]. Approval from the French regulatory body, Agence Nationale de Santé et du Médicament, and a national ethics committee to perform initial tests in vivo on healthy adult participants was granted in early 2023 (IDRCB 2022-A02321-42). Whether humans could withstand 1.5-hour MRI exams at 11.7T was uncharted territory, the preliminary magnet study (PREMS) therefore proceeded cautiously to gather evidence for the absence of harmful effects in realistic conditions [19, 20]. Physiological, vestibular, and cognitive measurements were recorded to verify the innocuousness of the magnetic field on 20 participants scanned at 11.7T, and on another 20 participants with the magnetic field switched off (nocebo effect). The participants of the OT control group were not informed of the absence of the magnetic field and received the same instructions as the volunteers scanned at 11.7T. The sound of MRI sequences was mimicked with recordings played by a loudspeaker hidden at the back of the magnet. Genotoxicity was assessed for the 11.7T group, with blood samples drawn before and after the 90-minute exposure. In all tests performed throughout the protocol, no significant differences between the 11.7T group and the OT group could be identified (for a comprehensive description and results of this study, see [21]).

Radiofrequency pulse design and parallel transmission technology

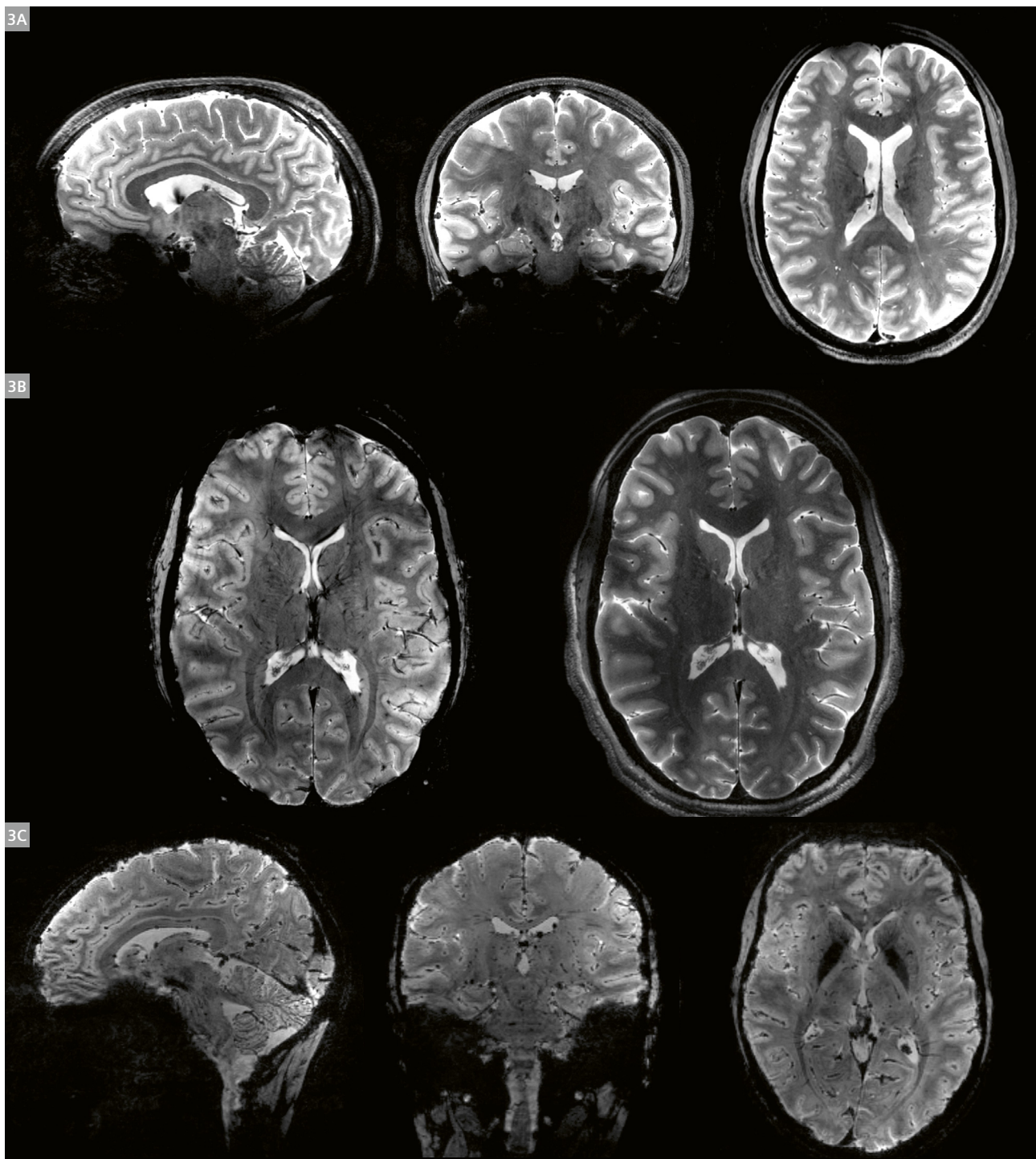
In MRI, the B_1^+ field refers to the radiofrequency (RF) magnetic field that is used to excite the protons in the body. Achieving homogeneous B_1^+ distribution in the body becomes increasingly challenging at higher field strengths due to the shorter wavelength of the RF field [22]. Destructive interferences lead to significant

spatial variations in the signal intensity, reducing image quality and causing artifacts in regions with complex geometries, like the human brain. One established method for mitigating these B_1^+ inhomogeneities is parallel transmission (pTx) technology, which uses multiple RF coil elements to independently control and modulate in time the amplitude and phase of the transmitted RF signals [23]. By employing pTx techniques, it is possible to tailor the RF fields to specific regions of interest, resulting in more uniform excitation and improved image homogeneity [24, 25]. With MAGNETOM Iseult, the use of pTx is mandatory to perform high-quality proton imaging of the human brain.

MR imaging at 11.7T

The first in vivo images of the human brain were acquired at 11.7T during the second half of 2023 on the PREMS cohort [21]. The magnetic resonance sessions were 90 minutes long and targeted anatomical T1-weighted, T2-weighted, and T2*-weighted contrasts, using gradient-echo (GRE) and spin-echo MR sequences (the main sequence parameters are provided in Table 2). The aforementioned homemade pTx RF coil with dedicated virtual observation points for specific-absorption-rate (SAR) monitoring was used for all imaging. All acquisitions were performed with up to second-order B_0 shimming using quadratic programming and brain masking. The RF and ΔB_0 field maps were acquired first for each participant for the pulse design and shimming using 2D interferometric saturated turboFLASH [26] at 5 mm isotropic resolution and 3D multi-GRE acquisitions at 2.5 mm isotropic resolution, respectively. RF pulses were designed with the k_r points [27], the “spokes” [28], and the gradient ascent pulse engineering (GRAPE) algorithm [29] approaches, respectively, using an active-set algorithm [30] with simultaneous k -space optimization [31] and under explicit hardware and SAR constraints. The GRAPE pulse was a universal RF pulse [32] and was designed offline on a database consisting of the first six participant field maps, while all other tailored RF pulses were designed using a dedicated workstation next to the MR system, during the exam. MR sequence source codes were modified to enable the use of the dynamic pTx RF pulses. No images were corrected for reception field bias (B_1^-). A gallery of images is shown in Figure 3 (to see more images, refer to [21], which is an open access article).

¹MAGNETOM Iseult is an investigational device only and is not intended for sale or clinical use. Research operations may require observation of national regulations.



3 In vivo images of the human brain obtained at 11.7T. **(3A)** 3D T2-weighted variable flip angle turbo spin-echo (SPACE) acquisition, with parallel transmission GRAPE universal RF pulses reaching a high RF excitation homogeneity throughout the whole brain. **(3B)** In vivo T2*-weighted 2D GRE axial image, with multi-spoke RF pulses reaching a high B1+ field homogeneity on a slice-by-slice basis [33], juxtaposed with T2-weighted 2D TSE axial images, obtained by combining two complementary RF static shims [34]. **(3C)** In vivo T2*-weighted 3D GRE images, used to perform quantitative susceptibility mapping (QSM) [35].

	2D T2* GRE	3D T2* GRE	2D TSE	3D T2 SPACE
TA (min)	5:16	5:14	4:26	10:08
TR (ms)	600	25	7000	6000
TE (ms)	20	3.5 / 8.12 / 12.74 / 17.36	48	301
FA (°)	27	10	180	T ₂ var
Matrix size	1032	276	640	400
Resolution (mm) ³	0.19 × 0.19 × 1	0.8 × 0.8 × 0.8	0.3 × 0.3 × 1	0.55 × 0.55 × 0.55
No. of slices/partitions	13	224	7	320
Acceleration	GRAPPA 2	GRAPPA 2×2	GRAPPA 4	GRAPPA 4×2
Turbo factor	–	–	9	165
Bandwidth (Hz/px)	40	430	130	250

Table 2: Main parameters of MR sequences used during the preliminary magnet study (PREMS).

A fruitful, ongoing collaboration between Siemens Healthineers and the CEA

The collaboration between the CEA and Siemens Healthineers in the field of MR essentially started with the first work packages for the Iseult project back in 2006, embodied by the signature of a master research agreement between the two entities. Siemens Healthineers has committed to providing MR instrumentation dedicated to this system, with the latest software and hardware technologies, bringing MAGNETOM Iseult to the highest standard of MR instrumentation from the outset. Several individual project agreements (IPAs) have also been signed to enable collaboration on specific topics. In particular, one IPA has been set up to develop and adapt a handful of state-of-the-art MR sequences provided by Siemens Healthineers to handle the specific pTx RF pulses designed using the tools of the CEA, so that they can be properly operated at 11.7T. Another IPA oversaw a collaboration to obtain key experimental data about the interaction between the gradient coil and the Iseult magnet to identify safe working conditions for using the MRI equipment.

Due to the magnitude of this project, people from different teams (project management, ultra-high-field development, MR collaboration) at Siemens Healthineers were involved. Customer service engineers were present at every stage of the installation, start-up, and operation of the MR system. The successful acquisition of the first in vivo images at 11.7T, with only 20 trials, is a perfect illustration of teamwork over many years and across the two organizations, with a great deal of preparation beforehand at 7T.

A promising future

The Iseult project is already 20 years old, yet somehow it feels as if this adventure is only just beginning. By producing the very first in vivo images of the human brain, this one-of-a-kind MR system has demonstrated great potential. While the PREMS study is now finished, more approvals from the regulatory body and national ethics committee are sought to perform new imaging sessions on healthy volunteers, particularly focusing on functional MRI (fMRI). Clinical research projects on patients are also envisioned. At this stage, many efforts are still ongoing to expand the current portfolio of pTx-compatible MR sequences, with more robust RF pulse design and additional sequence programming. For instance, designing robust inversion RF pulses was challenging due to the architecture of the present coil, and they left very localized artifacts despite achieving high overall performance [21]. More coils have also been designed and are ready to be tested [36]. One vision of CEA/NeuroSpin is to offer MAGNETOM Iseult as a unique international platform for the scientific community, making it accessible to any research project in the world, with dedicated on-site resources for support. Following the announcement of the first images, a number of sites have already approached the research center about launching new and ambitious clinical investigations.

Nevertheless, many of the challenges faced are further amplified at higher magnetic field strengths. The high spatial resolution aimed for makes imaging far more susceptible to blurring, and nearly two-thirds of the images acquired during the PREMS cohort had to be discarded because of motion artifacts. To tackle this wide range of challenges, the NeuroSpin team is working within the European Accurate, Reliable and Optimized Functional Magnetic Resonance Imaging (AROMA) consortium (<https://aroma-h2020.com/>) to assemble the necessary international expertise.

Increased static field inhomogeneities also are a burden for echo-planar imaging, the cornerstone of functional exploration of the human brain, causing unwanted image distortions and signal losses. A new multi-coil array prototype, named SCOTCH [17], has recently been designed by the CEA to shim the static field in the brain at 7T as if it were a partial seventh-order spherical harmonic conventional shimming system. Simulations showed promising artifact reduction at 11.7T with a 48-channel SCOTCH with a higher current amplitude. However, it would require a high-performance cooling system to dissipate the power. The implementation of such system and its positive impact on functional MR image quality is the subject of current research being carried out at CEA/NeuroSpin.

A central goal is to push the resolution of human fMRI to new levels, for instance by resolving individual cortical layers with spatial resolution on the order of 350–450 micrometers [4]. In view of the hardware needed to achieve such an ambitious target, MAGNETOM Iseult will be upgraded in the coming years with a powerful new gradient head coil designed by Siemens Healthineers (called Impulse, it is capable of achieving a maximum gradient strength of up to 200 mT/m for each axis and a slew rate of up to 700 T/m/s for each axis) and a state-of-the-art RF high density receiver array [4] that allows the system to quickly pick up fleeting MR signal before it decays.

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A Milestone in Medical Imaging: The Pioneering Development of the First Integrated PET/MR Scanner at Siemens Healthineers

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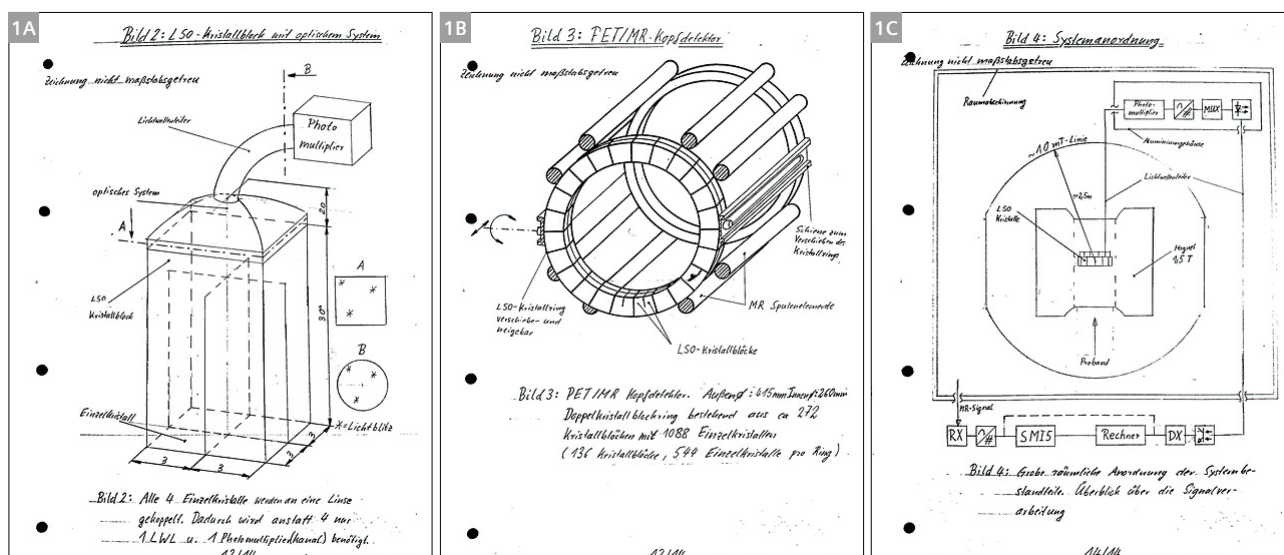
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Introduction

Siemens was the first company to build and market a whole-body PET/MR for clinical imaging. The story of this landmark achievement is worth retelling. PET/MR is a powerful imaging technology that uses the strengths of both modalities synergistically, combining the submillimeter, anatomic resolution of MRI with the ultra-high sensitivity of molecular PET imaging. Due to the complementary nature of the information provided by each modality, PET/MR is ideal for complex disease characterization and radiomics. In contrast to PET/CT, PET/MR also allows for truly simultaneous and isocentric imaging. Therefore, fully

synergistic applications like MR-guided PET motion correction and PET-guided MR motion correction are also possible. In addition, the technology allows multifunctional imaging, i.e., the combination of functional information derived from MR and PET, such as BOLD fMRI with ¹⁸F-FDG PET [1].

Compared with PET/CT, which was already clinically available in 2000 [2], PET/MR had a slower start, with the first human images being published in 2008 [3]. This was mainly due to the high complexity and technical challenges of combining both modalities into a single system.



1 PET/MR idea from 1998. (1A) Shows the LSO crystals coupled via a light guide to a fiber-optic cable and PMTs outside the magnetic fringe field. (1B) The system was envisioned as a brain-PET detector with a movable detector ring for FOV selection. (1C) System layout with electronics installed inside and outside of the RF cabin. PET/MR image fusion was envisioned to be performed on an extra computer coupled with the Siemens Medical Imager Five (SMI5) image reconstruction computer of the MR system.

In addition, while PET/CT simplified the workflow of a PET exam by reducing the time to acquire an attenuation correction for the PET reconstruction, PET/MR first needed to establish applications, workflows, and protocols – some of which incorporate options unique to PET/MR.

This article aims to reconstruct part of the history of the technology from the point of view of Siemens Healthineers. It is not intended to give a full and exact account of the development of PET/MR itself, which has been the subject of considerable research and development work over the past 30 years.

Initial idea to combine PET and MR imaging

Engineers and physicists at Siemens Medical Solutions, as our company was then called, of course closely monitored the technological developments in both modalities as part of their daily routine. In the mid-to-late 1990s, the academic world was also studying the foundations of future PET/MR systems in the form of photomultiplier tube (PMT) detectors coupled with optical fibers [4], and semiconductor-based light detectors such as avalanche photodiodes (APDs) [5]. As early as 1998, the first idea for a combined PET/MR system was presented within Siemens. A system was described that operated in the form of a head coil with add-on PET detectors (Fig. 1), connected by optical fibers or APDs. Since 3T was by no means an established field strength in the clinical field, the improvement idea referred more specifically to systems like the 1.5T MAGNETOM Vision or one of its successors, such as the 1.5T MAGNETOM Symphony or the 1T MAGNETOM Harmony. The idea was ahead of its time, and the proposal was rejected. Nevertheless, the concept of PET/MR made a successful return within 10 years.

Testing the waters: The BrainPET system

By the mid-2000s, it was evident that PET/CT systems were often superior to standalone PET systems in clinical practice due to the combination of patient throughput, of physiological information and of anatomical detail. However, in certain applications, such as neurology and oncology, MRI [6] showed clear advantages over CT imaging due to its superior soft-tissue contrast and capabilities for functional imaging, such as blood-oxygen-level-dependent (BOLD) imaging. Small-animal PET/MR systems showed that it was possible to solve the basic technical challenges, such as crosstalk between the two modalities [7–9]. Therefore, the idea of a combined PET/MR system regained traction at Siemens. In 2005, Siemens Medical Solutions bought its longtime PET partner, CTI Molecular Imaging, with the goal of expanding its presence in the molecular imaging market [10]. Pioneering efforts to build solid-state detectors for PET/MR

with APDs were already underway at CTI, so it was agreed to go forward with a combined PET/MR system to see if a suitably large PET/MR market existed. There was a clear vision to work toward a whole-body PET/MR system, but due to the many technical challenges and the disruptive nature of the system, it was decided to take an intermediate step via a research system to learn, and then to iterate from there.

The system later became known as BrainPET, which reflected its main application. Conventional PET systems at this time still used PMTs for photodetection. However, PMTs are very sensitive to external magnetic fields, rendering them unusable in the vicinity of an MRI system. Using long optical fibers to connect the scintillation crystals to the PMTs was deemed inefficient due to inherent light loss and signal shape changes. Fortunately, APDs were just becoming commercially available and were stable enough. They were used in the calorimeter at CERN (The Electromagnetic Calorimeter Project, Technical Design Report, CERN/LHCC97-31, 1997), so the Siemens developers knew that Hamamatsu APDs worked and were stable. In addition, preclinical systems existed [5]. The form factor and resolution of the APDs offered the possibility of resolving smaller crystals. It was therefore a natural step in detector development to move on to the APD technology. Ultimately, this was also a preliminary development for more advanced solid-state light detectors such as silicon photomultipliers (SiPMs), though these were not ready to use at that time.

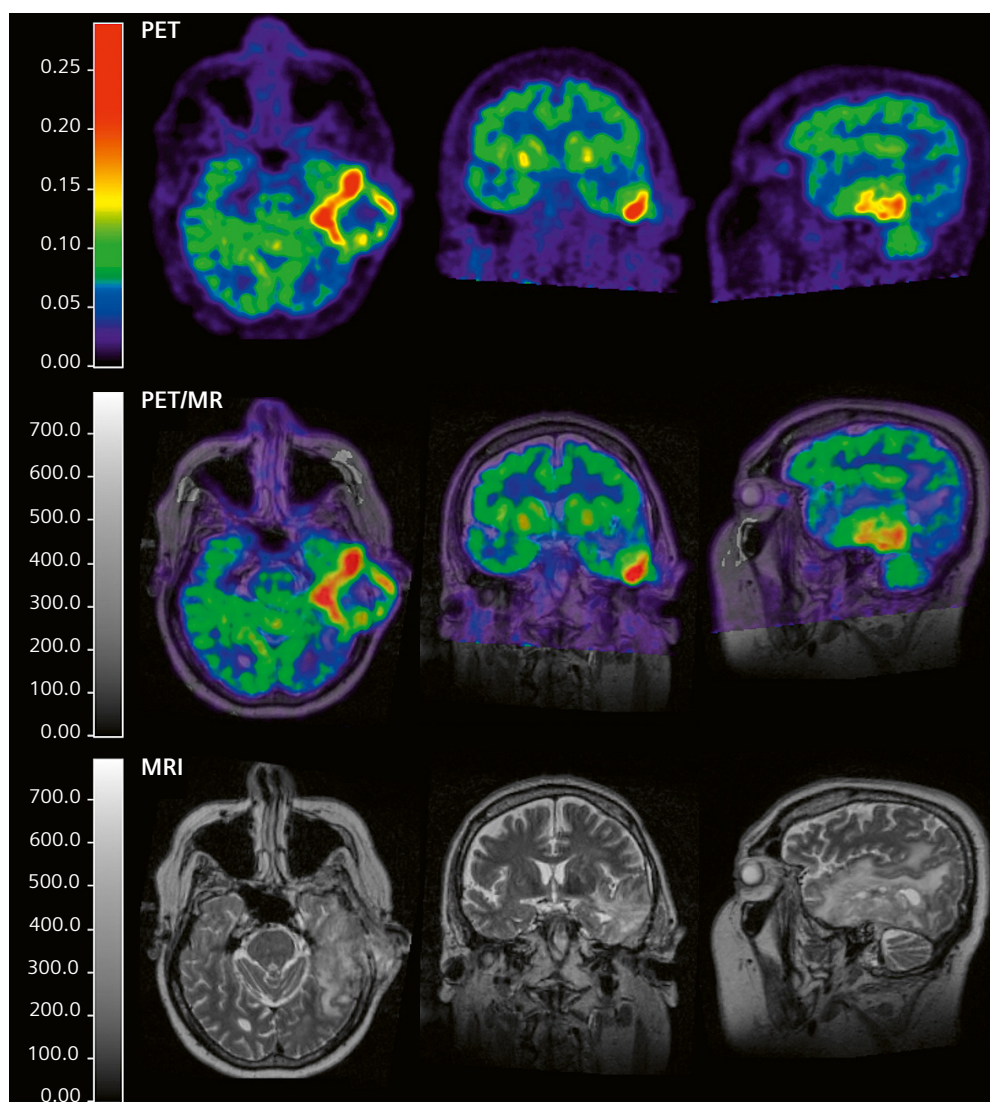
Therefore, the system used CTI's newly developed PET detector, for which the pioneered APDs were coupled to a matrix of $2.5 \times 2.5 \times 20 \text{ mm}^3$ lutetium oxyorthosilicate (LSO) scintillation crystals. Six of these modules were combined into a detector cassette enabling a field of view (FOV) of 19.2 mm in the z-direction (axial). Thirty-two of these copper-shielded cassettes formed the cylindrical PET detector system. The inner diameter of the PET detector was 36 cm, with the outer diameter fitting inside the 60-cm bore of a 3T MAGNETOM Trio MRI scanner. To enable MR signal transmission and reception, a dedicated transmit receive coil was fit inside the PET ring – a birdcage transmission coil and an eight-channel receive coil.

This combination essentially transformed the whole-body MAGNETOM Trio MRI system into a head-only (and some body parts) PET/MR system [11]. The PET insert was completely retractable (although its cables were somewhat heavy) and, without the insert, the system could be used as a regular clinical MRI scanner. The PET detector cassettes were connected to the Siemens Medical Solutions QuickSilver data acquisition system [12] via 10-meter cables. A PET handling device mounted at the end of the patient bore allowed the setup of a PET/MR system or an MR-only system. The BrainPET prototype was designed mainly to gain experience of this new modality

and was therefore introduced at five sites. These were: Atlanta, GA, and Boston, MA, in the United States; and Jülich, Munich, and Tübingen in Germany. Two of these BrainPETs are still used for research in Boston and Jülich. Their image resolution makes them particularly attractive for neurological research. With the introduction of the BrainPET, Siemens Medical Solutions established a strong collaboration network to further develop PET/MR technology. Amongst the big successes of BrainPET were the first human PET/MR images obtained by a U.S.-German team in 2008 [3], and extensive and high-ranking research work published and presented by the respective sites [13–17]. Figure 2 depicts one of the first PET/MR images acquired on the BrainPET: the first patient scanned at the Athinoula A. Martinos Center for Biomedical Imaging on May 23, 2008. The patient was a 54-year-old man with glioblastoma with cutaneous extension and received 5.45 mCi of FDG for a clinical PET scan performed 2.5 hours before the PET/MR scan.

The BrainPET sparked many creative research projects and academic ambitions, and the nuclear-medicine and MRI communities were very positive about the marriage of these two modalities. However, the limitations of a dedicated system for neurology research were becoming obvious, as it was not intended as a PET/MR system for widescale use. Huge potential was predicted for the fields of oncology and cardiology, both of which would require a whole-body PET/MR system. Furthermore, the BrainPET was not yet a fully integrated system: The MR and PET images had to be acquired and reconstructed at two separate consoles and there was no integrated solution for attenuation correction.

Still, the BrainPET allowed the collection of important clinical and technical experience and feedback, enabling PET/MR development to take the next quantum leap toward a fully integrated, clinical whole-body system. This became known as the Biograph mMR.



2 First simultaneous ^{18}F -FDG PET/MR scan at the Athinoula A. Martinos Center for Biomedical Imaging using the 3T MR-compatible BrainPET prototype. The subject was a 54-year-old man with glioblastoma with cutaneous extension. Images courtesy of Ciprian Catana, MD, PhD; A.A. Martinos Center, Massachusetts General Hospital, Boston, MA, USA.

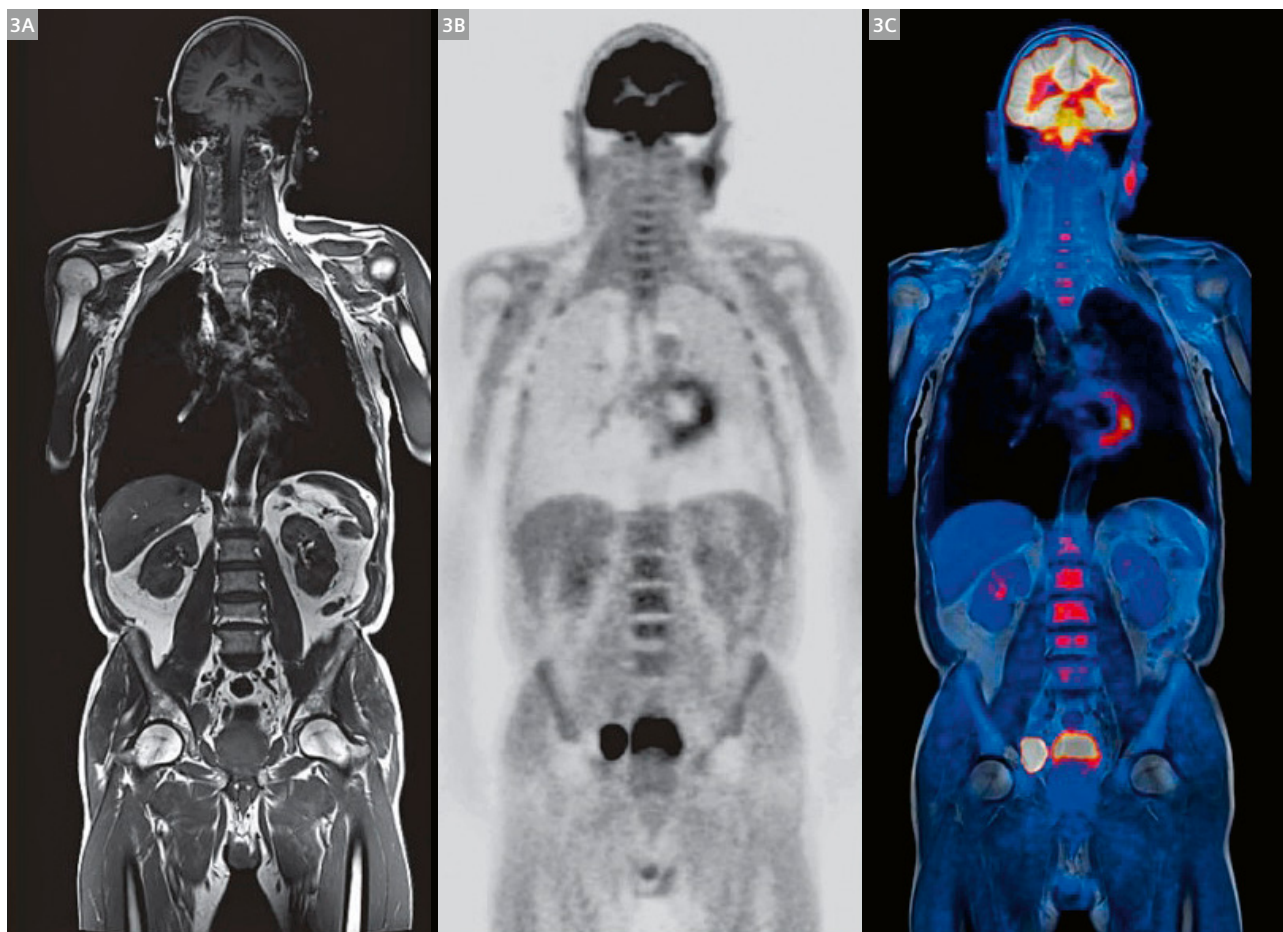
From simultaneous to synergistic PET/MR: Biograph mMR

Siemens had developed concepts for a whole-body all-purpose scanner in parallel with the introduction of BrainPET and maintained this vision in ongoing dialog with the PET/MR clinical and research communities. Various challenges had to be overcome on the electronics and mechanical integration front.

In the transition from BrainPET to the Biograph mMR, APDs were still used, but the electronics were completely redesigned so they could sit inside the main magnetic field, inside the bore or near the end of the magnet rather than outside the main magnetic field. This forced further development in digitalization, including digital time measurement of PET annihilation photon events directly at the MRI scanner, which was a considerable innovation at the time.

The mechanical integration concept of the PET gantry into the MRI system was also a challenge. This included: the ability to service the PET subsystem in the magnetic field without disassembling the unit; an interleaved cooling concept; mutual shielding of each modality from the adverse effects of the other; a holistic concept for attenuation correction for both the patient, the patient table, and the MR coils; plus additional topics regarding hardware and software integration, including temporal and spatial registration of the acquired data and images.

It became clear early on that the whole-body PET/MR system had to be based on an MRI system with a 70-cm bore, and that the PET detector needed to fit within the MR scanner and have a field of view that did not extend out of the homogeneous field of the MR, to make an integrated solution possible.



3 Patient with lymph node metastasis following a history of a penis carcinoma. Images show whole-body T1-weighted MR image (3A), attenuation corrected PET (3B), and MR/PET hybrid images (3C) in a coronal orientation, that were all simultaneously acquired in a multistation/multi-bed acquisition mode. MRI in this wholebody study (3A) displays anatomical structures with exquisite detail and excellent soft-tissue contrast. PET (3B) and the combined MR/PET hybrid image (3C) in this coronal overview show enhanced tracer activity in one lymph node on the right side of the patient.

Images from Quick HH, Ladebeck R, Georgi JC. Whole-Body MR/PET Hybrid Imaging: Technical Considerations, Clinical Workflow, and Initial Results. *MAGNETOM Flash*. 2011; 46(1):88–100.

Numerous discussions were also held regarding the MR field strength of such a system. The clinical standard was 1.5T, but 3T was the field strength preferred by researchers and clinicians with a focus on neurological imaging. Given the experiences and positive feedback from BrainPET users, the decision was made in favor of a 3T-based system. It is worth mentioning that Siemens conceived this system for both clinical and research use from the outset. The specification of the system was designed to compete with the high-end systems of each modality, and to allow for optimal clinical performance and excellent research. The PET detector of the Biograph mMR system was based on APD electronics and $4 \times 4 \times 20$ mm LSO crystals; 8×8 crystals were coupled with 3×3 APDs on a detector block. Eight rings each with 56 detector blocks formed the PET detector with an axial FOV of 25.8 cm (head-foot direction). The integration of the detector reduced the bore size from 70 cm to 60 cm. The motivation was to enable the growing PET/MR community to pave the way for widespread PET/MR usage through pilot studies and by identifying the key advantages and most compelling clinical applications. The first whole-body scan was performed during a clinical investigation at the first external installation at the Technical University of Munich, Germany, in November 2010.

When the Biograph mMR was launched at the RSNA in 2010 and with the commercial system installation in 2011, it was a revolution. People were excited and could barely believe it was an integrated system with all the capabilities, but the small footprint of a regular MRI system. According to one anecdote, people walked past the conference booth wondering where to find the integrated PET/MR system, because the Biograph mMR looked just like a regular MRI scanner from the outside. Another anecdote shows the excitement that the Biograph mMR generated everywhere: For the first time in the history of Siemens Medical, the FDA's press statement saying the system was cleared for the U.S. market was issued *before* the Siemens marketing department was able to publish its own.

The progress of the Biograph mMR since 2011 can roughly be divided into three phases: During the first phase the installed base grew quickly. Most of the systems were installed at sites which used the system for research and where many innovative ideas were tried. Some proved successful, others were discarded. The community invested a lot of effort in proving and discussing equality with respect to PET/CT. Technical developments improved the performance of the system and led to the introduction of a new software version in 2014. A range of applications and protocols were proposed and tried. Additional features (truly synergistic applications, e.g., respiratory motion management and respiratory gating both with a matching attenuation correction) were introduced along with a first attempt at MR-driven PET head motion correction

(BrainCOMPASS, with NEMA and ACRIN testing capabilities) and more.

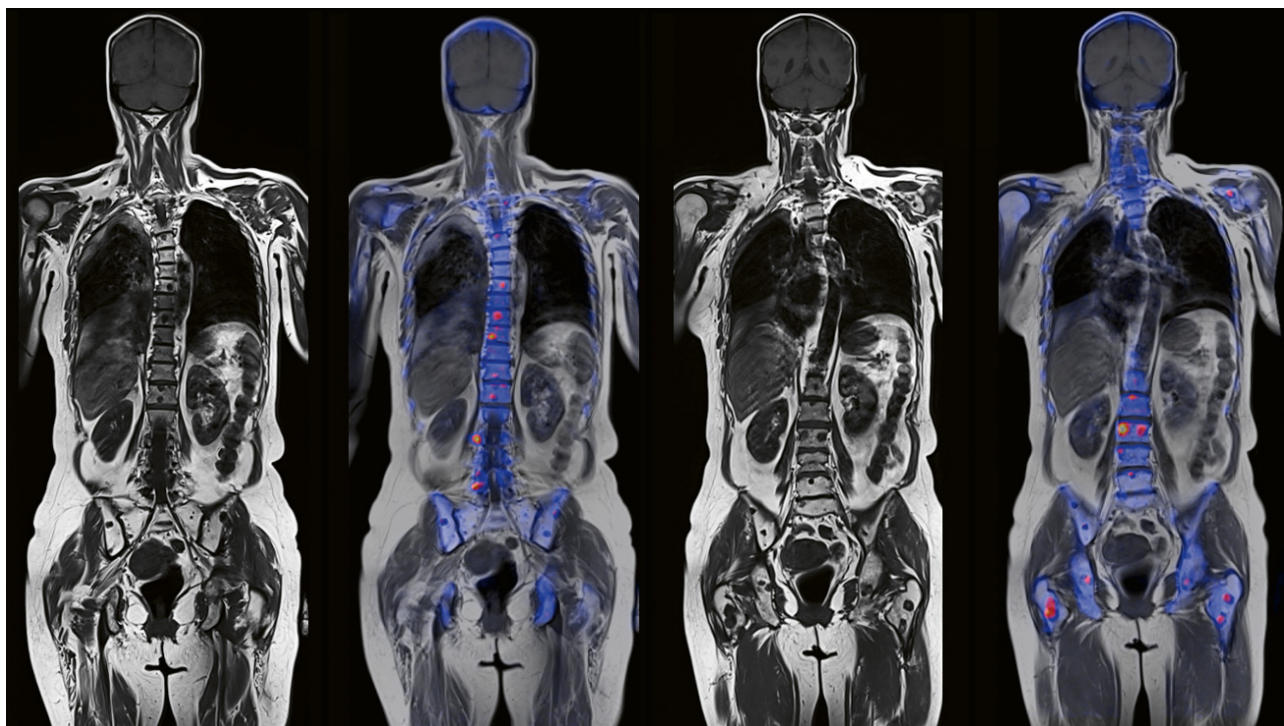
In the second phase specific key applications emerged, e.g., indications for which a PET/CT and an MRI scan are generally ordered (epilepsy surgery planning, dementia evaluation, oncology staging, prostate cancer staging, neurodegenerative disease, cardiac sarcoidosis). Another important aspect was the reduction of radiation exposure from the CT component, making it particularly suitable for pediatric malignancy cases [18, 19]. Protocols were becoming more streamlined, and consensus papers on PET/MR protocols were published [20]. A new software version was introduced in 2017 with further improvements in attenuation correction (extended FOV, and bone attenuation correction) and truly synergistic motion correction (BodyCOMPASS), as well as improvements to the general PET/MR workflow (Fig. 4).

Roughly since 2017, patients with these key indications are being scanned in financially sustainable volumes, and new tracers are boosting PET/MR usage (tau- and amyloid-based tracers for Alzheimer's diagnostics, and PSMA for prostate imaging). Research is continuing extensively. The community is requesting further innovations in the areas of PET and MRI such as imaging pain or using MRI to anatomically guide the PET reconstruction of brain and prostate images.

For three or four years, the Biograph mMR was the only clinical whole-body PET/MR scanner on the market. Even 13 years after its introduction, it has retained its attractiveness as the most complete and mature commercial PET/MR system in the world. So far, an estimated 250,000 patients have been scanned on Biograph mMR systems, meaning a PET/MR system has helped write a quarter of a million personal healthcare stories. Many case studies and reports have been published [21–23] describing how complementary information acquired with PET and MRI has made a difference to the patient's diagnostic outcome and the chosen treatment path.

Shape the new era of personalized care: BIOGRAPH One

Since the introduction of Biograph mMR in 2011, the PET/MR market has experienced steady growth. Initially, the focus was primarily on research, but as the installed base expanded, clinical applications began to emerge, particularly in areas where the combined strengths of soft tissue contrast and resolution from MRI and metabolic insights from PET provided unique diagnostic value. Key application areas include neurology (e.g., neurodegenerative diseases), oncology (e.g., prostate cancer imaging), cardiology, and theranostics, with uses in oncology, inflammatory diseases, and novel Alzheimer's treatments.



4 Metastatic breast cancer ^{18}F -NaF PET/MR (Biograph mMR). A patient with bone metastases demonstrates the potential of the Biograph mMR to help in the diagnosis of bone processes either osteoblastic, osteolytic or infiltrative. Fat suppressed MRI is advantageous to detect early marrow changes. PET overlapped with T1 TSE composed. *Images courtesy of NYU Langone Medical Center, New York, USA.*

Despite these successes, wider adoption of PET/MR was hindered by three main challenges identified a decade after its market introduction:

- **Economic barriers:** Longer scan times, higher costs compared to PET/CT, and limited reimbursement options.
- **Complexity:** Complex workflows, extensive patient preparation, high data volumes, and lengthy reading times.
- **Technological gaps:** Image resolution and contrast lagged behind the capabilities of the latest standalone PET and MRI systems.

To address these challenges, Siemens Healthineers leveraged flagship technologies from both PET and MRI modalities and added significant innovations. A great example is the PET detector where the novel approach to digitalize the signals already on the detector board level, all components except the LSO crystals and the SiPMs had to be newly designed. Thus developing the next-generation system, now introduced as BIOGRAPH One¹. This new system is designed to integrate key technologies from the BIOGRAPH Vision PET/CT and the MAGNETOM Vida MRI scanner, while adding numerous innovations on top addressing the following:

Designed to promote high patient throughput:

- Boost PET performance by introducing SiPM technology and best-in-class time-of-flight (TOF) imaging
- Boost MRI performance by introducing deep learning image reconstruction
- Provide high-density channel MRI coils to allow for higher acceleration factors
- Extend the PET FOV to 35 cm to reduce the number of bed positions in whole-body acquisitions and to increase the PET sensitivity
- Improve the up time of the system by reducing the time for system calibration and detector replacement

Reduce system complexity:

- Introduce a new workflow to simplify planning and scan execution, bringing components from MR and PET/CT planning, in a unified user-interface
- Introduce protocol automation such as myExam Assist, AutoAlign, and the copy reference concept to reduce exam planning and user interaction and to improve workflow consistency
- Improve serviceability and operational aspects of the scanner, such as reducing siting requirements by utilizing only one filter plate

¹BIOGRAPH One is currently under development and not commercially available. It is not for sale in the U.S.A. and its future availability cannot be ensured.

In the PET domain, the introduction of TOF technology and SiPM detectors has significantly enhanced imaging performance. By 2017, the BIOGRAPH Vision PET/CT was introduced, as one of the first in the industry to offer a TOF timing resolution below 250 ps, reducing time resolution by 50% compared to other PET/MR systems on the market. This leap improved effective sensitivity and localization accuracy – critical for clinical applications, especially in neurology. Additionally, the shift from analog to digital detectors reduced signal interference and increased overall robustness. These advancements along with additional key innovations, are now first introduced with BIOGRAPH One, designed to result in an order of magnitude improvement in PET sensitivity.

On the MR side, BIOGRAPH One is planned to incorporate the latest 3T technology with BioMatrix technology, advanced coil designs, and cutting-edge reconstruction methods like Compressed Sensing and deep learning-based image reconstruction (Deep Resolve). These upgrades enhance image quality, accelerate scans, and increase patient throughput, making the system well-suited for both clinical and research applications.

The development of BIOGRAPH One aimed to overcome economic and operational barriers, while also boosting clinical performance. Enhanced PET sensitivity is achieved through SiPM technology and TOF imaging. Deep learning-based image reconstruction improves MR image quality, and the extended PET field-of-view (FOV) reduces the number of bed positions required for whole-body scans. Improved calibration and streamlined maintenance processes further enhance system uptime and reliability.

The introduction of a unified user interface and automation tools such as myExam Assist and AutoAlign will simplify workflow, reducing the time needed for exam planning and improving consistency. By integrating these elements, BIOGRAPH One offers potential for a seamless,

simultaneous PET/MR experience that addresses the demands of modern clinical environments.

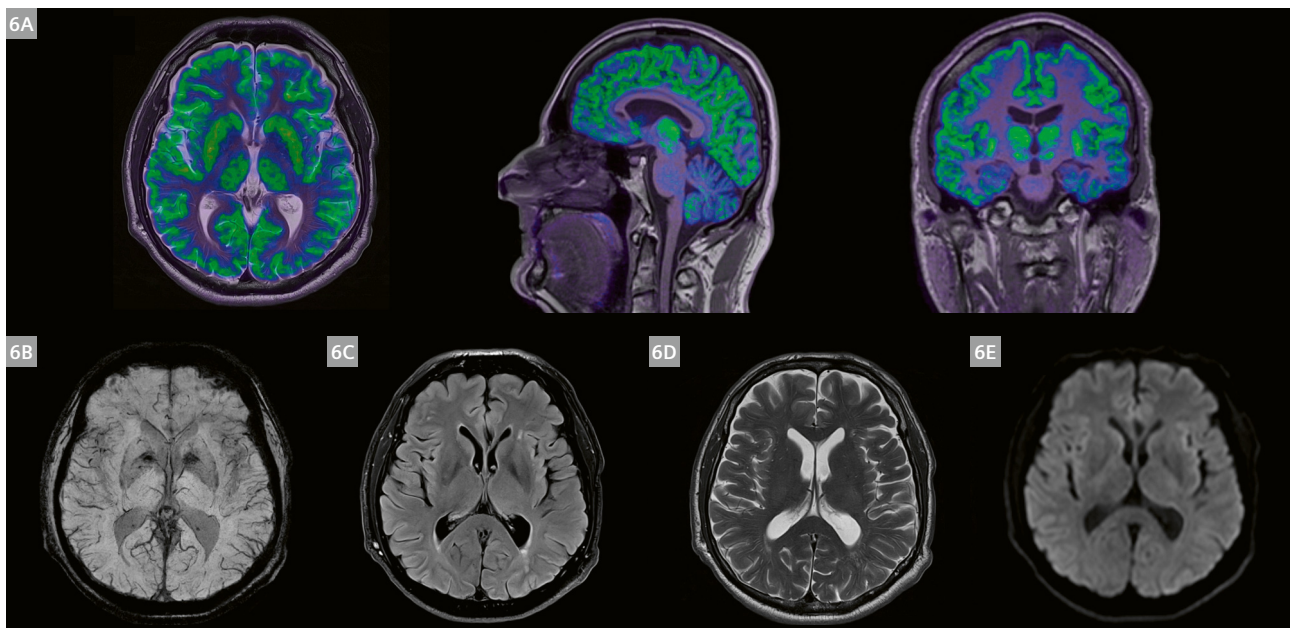
The journey of BIOGRAPH One has reached an exciting new chapter with the first system installation in Rigshospitalet Copenhagen, Denmark (Fig. 5). The very first clinical scans were conducted as part of the system's initial evaluation. These included a scan of a patient undergoing assessment for suspected dementia (Fig. 6) and another of a patient with relapsed follicular lymphoma being evaluated for complications related to immunosuppression and chronic lung disease (Fig. 7). These are some of prime clinical applications of PET/MR. This is just the beginning – with more installations to follow, we look forward to continuing Siemens Healthineers' leadership in PET/MR imaging and shaping this new era in personalized care with BIOGRAPH One.

The evolution of PET/MR has not only advanced diagnostic capabilities but has also broadened the horizon for clinical and research applications. With every iteration, the technology has shifted closer to fulfilling its potential: delivering a comprehensive, efficient, and user-friendly diagnostic tool that combines the best of PET and MR into a single, seamless experience. Siemens Healthineers' dedication to overcoming technical and economic barriers continues to drive innovation, and the BIOGRAPH One stands as a testament to this ambition.

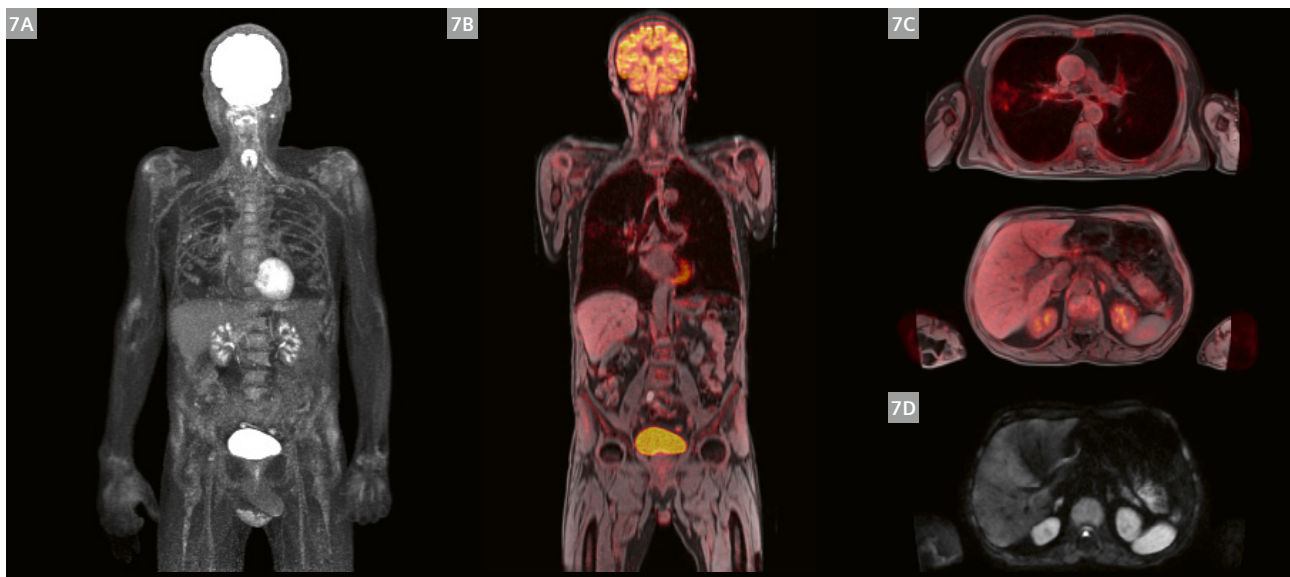
By blending cutting-edge developments in PET sensitivity, MR efficiency, and user workflow, Siemens Healthineers reaffirms its role as a pioneer in the imaging industry. The future holds new possibilities for PET/MR, promising expanded applications, improved patient outcomes, and a continued commitment to pushing the boundaries of what hybrid imaging can achieve. This new era of PET/MR will not only serve the needs of today's clinicians and patients but will lay the foundation for tomorrow's breakthroughs in precision medicine.



5 The first BIOGRAPH One. Assembled in the Siemens Healthineers headquarters in Erlangen, Germany and installed at University Hospital Erlangen, Germany, and at the Rigshospitalet, Copenhagen, Denmark.



6 First-ever clinical images acquired using the new BIOGRAPH One system as part of the clinical investigation being conducted at the Rigshospitalet, Copenhagen, Denmark. A ^{18}F -FDG PET/MR scan was performed on a 63-year-old patient, who presented a history of progressive memory issues and was being assessed for suspected early neurodegenerative disease. The exam revealed slightly asymmetrical uptake in the frontal regions, with preserved cortical metabolism and symmetrical uptake in the basal ganglia, thalami, and cerebellar hemispheres. **(6A)** Fused PET and T2-weighted anatomical MR images displayed in axial, sagittal, and coronal views. **(6B)** Susceptibility Weighted Imaging provides high sensitivity for detecting vascular structures and micro-hemorrhages. **(6C)** T2-weighted Dark Fluid sequence enhances contrast for cerebrospinal fluid spaces, aiding in visualization of potential abnormalities. **(6D)** High-resolution T2-weighted anatomical scan, used for detailed structural assessment. **(6E)** Diffusion-weighted imaging at $b = 1000 \text{ s/mm}^2$, offering insights into tissue microstructure without specific abnormalities detected.



7 First clinical images from the BIOGRAPH One PET/MR system at Rigshospitalet, Copenhagen, Denmark. A 75-year-old male patient with relapsed follicular lymphoma was scanned to assess immunosuppression-related complications, including recurrent pneumonia and Chronic Obstructive Pulmonary Disease (COPD). The imaging revealed unchanged metabolic remission of lymphoma but increased FDG uptake in the lungs, consistent with inflammatory or infectious changes. **(7A)** PET-only image showing metabolic activity in the lungs and parotid gland. **(7B)** Fused PET and MR attenuation-corrected image (PET+MRAC) for anatomical localization and **(7C)** axial views of increased FDG uptake. **(7D)** Diffusion-weighted imaging (DWI) sequence, providing additional information on tissue composition and potential microstructural changes.

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From left to right: Marc Lauterbach, Silvia Dias Almeida, Hans Wehrl, Holger Schmidt, Manuel Schneider. Not in the picture are Matthias Schmand, James Corbeil, David Faul, and Matthias Fenchel

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Meet Siemens Healthineers

Siemens Healthineers: Our brand name embodies the pioneering spirit and engineering expertise that is unique in the healthcare industry. The people working for Siemens Healthineers are totally committed to the company they work for, and are passionate about their technology. In this section we introduce you to colleagues from all over the world – people who put their hearts into what they do.

Vibhas Deshpande, Ph.D.

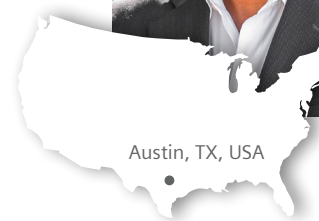
Vibhas Deshpande is Head of Sustainability Innovation and Strategic Research, and a Principal Key Expert at Siemens Healthineers North America. In this leadership role, Vibhas spearheads the development of solutions aimed at reducing the environmental footprint of healthcare while driving research to generate value from cross-modal, disease-focused technologies in healthcare diagnostics.

Vibhas leads cross-organizational innovation teams, influencing design improvements in next-generation medical imaging equipment and bringing sustainable healthcare solutions to market. He also shapes policy discussions with actionable insights, education, and data that advance the sustainability goals of healthcare systems worldwide.

Among his notable achievements, Vibhas conceptualized and drove the ideation and continued realization of the first carbon-neutral radiology imaging service in partnership with the University of California, San Francisco in 2021. This groundbreaking initiative was recognized with the Siemens AG U.S. Excellence Award in 2023, as one of the company's most impactful projects. Also in 2023, the Academy for Radiology & Biomedical Imaging Research named Vibhas a distinguished investigator for his contributions to biomedical imaging and healthcare innovation. Vibhas is the recipient of three active NIH grants for research in liver disease and carotid disease diagnoses. He is also an inventor on 21 granted patents, and has written four book chapters and over 55 peer-reviewed publications.

Vibhas began his career with Siemens Healthineers in 2003 as a senior scientist in MRI research and has held various roles in R&D and leadership, focusing on advancing medical technologies. He holds a BSc in electrical engineering from Savitribai Phule Pune University in Maharashtra, India, and a PhD in biomedical engineering from Northwestern University in Evanston, IL, USA.

Vibhas lives in Austin, Texas, with his wife and two daughters.



How did you first come in contact with MRI?

My first connection with MRI was a class I took in medical imaging during my PhD studies at Northwestern University. I found MR physics absolutely fascinating. I imagined that working in the field would let me help patients have more positive health outcomes, so it seemed like the perfect promise of academic and aspirational fulfilment. I decided to spend three months in the MRI research lab to find out if it was what I wanted to pursue as a career. My first day in the lab was mostly orientation, meeting the researchers

and support staff, and attending a study at the scanner. It sounded benign, but little did I know what was to come. Minutes into my introduction to a real-world clinical MRI scanner, the patient had an allergic reaction to the contrast agent, and... well, I'll leave the remaining description to your imagination. I wonder who was more traumatized that day: the subject or I 😬? I survived the experience and fortunately stuck around. Over 20 years later, here I am, still working on MRI and other healthcare innovations and looking forward to being here for many more years!

What do you find motivating about your job?

This may be a cliché, but for me, the timeless poem “What is Success” (said to have been written by Bessie Anderson Stanley in 1904, although it is frequently misattributed to Ralph Waldo Emerson) has always resonated (that word choice is not at all influenced by the MR physicist in me 😊). Working in medical imaging for many years, there have been numerous occasions when I have experienced first-hand how the work that imaging researchers do impacts decisions for a patient’s care and well-being. Being fortunate enough to have the opportunity to make a positive impact—however small, directly or indirectly—on anyone’s life, especially when they are at their most vulnerable, is a bottomless source of motivation. I am immensely grateful that this is supported and encouraged by Siemens Healthineers—an organization that whole-heartedly aims to provide healthcare for everyone, everywhere, with responsible environmental stewardship, and colleagues and customers who are extremely collaborative, open to innovation in every aspect, and willing to go the extra mile.

What are the biggest challenges?

We face numerous major challenges today in public health. These include a lack of access to healthcare, rising costs of healthcare, and the inequities in healthcare outcomes for vulnerable populations. The sustainability and resilience of a healthcare system are very complex issues that are influenced by, among other things, geopolitics, government policies, changing demographics, healthcare workforce shortages in many parts of the world, market dynamics, and climate change. A healthcare system transformation is needed to address our health needs in the decades to come. While technological innovation is a critical part of this transformation, the urgent need for all the vectors I mentioned above to align is the biggest challenge in my opinion. This same challenge of stakeholder alignment presents itself in different forms in every project, large or small. Communicating effectively and integrating different perspectives are often daily challenges, but they are also our biggest opportunities.

What are the most important developments in healthcare?

I am always fascinated by looking back at what have proven to be the most impactful interventions for human health: antibiotics, vaccines, anesthesia, diagnostic tools, genomics, and many more, are huge scientific achievements and relatively well-known. But what is striking to me is that providing access to clean water is consistently ranked as one of the most impactful public health interventions in human history. Clean water, something so basic that most of us likely take it for granted! Programs in the past decades enabling access to clean water around the world have reduced premature mortality, improved child health, and provided economic and social benefits to families.

While there are several important developments going on in healthcare today, I find two broad trends particularly inspiring: personalized medicine, and the early stages of a transformation toward greater preventive care in health systems.

The progress in personalized medicine is a crucial development, whether it is achieved through one or a combination of the following, to name just a few: artificial intelligence; genetics; epigenetics; digital twins; an improved understanding of mental health; a systems-thinking approach to disease; new, improved, and personalized therapies for cancer and chronic health conditions; and work on integrating the social determinants of health into care pathways.

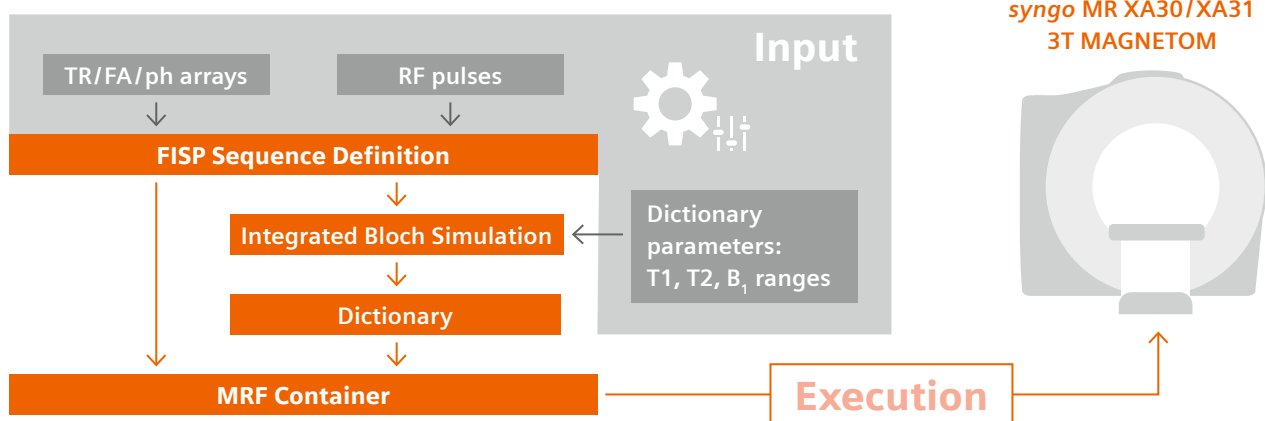
Orienting the health system toward preventive care by providing integrated healthcare is a very promising direction for improving outcomes with easier access to the health system, reduced costs, and improved health equity.

What would you do if you could spend a month doing whatever you wanted?

I very much enjoy traveling with my wife, children, and other family and friends. Hiking in Patagonia and climbing Mount Kilimanjaro are at the top of my list today. What I enjoy most when traveling is not only experiencing the magnificence of nature, but also meeting new people and listening to their life stories, always such a world away from my own!

MR Fingerprinting Development Kit

for Tailored Research Applications



The MR Fingerprinting Development Kit¹ takes files describing the input parameters T1, T2, RF- flip angles and pulse phases. This input generates a customized MR Fingerprinting¹ sequence. The input also serves as the basis for a Bloch simulation, which creates a MR Fingerprinting dictionary unique to its input parameters.

In a final step, the new MR Fingerprinting sequence and its unique dictionary are bundled in the so-called MR Fingerprinting container. This container can be transferred and executed on a qualified MAGNETOM 3T scanner from Siemens Healthineers² with a valid MR Fingerprinting license.

The sequence generation, the Bloch simulation, and the creation of the MR Fingerprinting container are performed by running a Windows command line tool³ provided with the MR Fingerprinting Development Kit.

The MR Fingerprinting Development Kit is available for download to all users of MR Fingerprinting at www.magnetomworld.siemens-healthineers.com/hot-topics/mr-fingerprinting/mrf-developer-kit

This unique MRF Development Kit empowers you to tailor the research application to your needs.

¹The product / feature is not for sale in the U.S. Its future availability cannot be guaranteed.

²syngo MR XA30: MAGNETOM Prisma, MAGNETOM Prisma Fit, MAGNETOM Skyra; syngo MR XA31: MAGNETOM Vida

³The user needs to install the MR Fingerprinting Development Kit on a standalone Windows 10 PC.

Installation on the host computer is not possible.

The entire editorial staff at Seoul National University Hospital, Seoul National University College of Medicine and at Siemens Healthineers 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.

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