

MAGNETOM Flash

Issue Number 98 · 3/2026

ISS Edition

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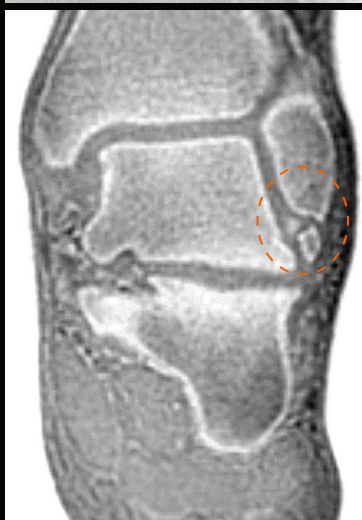
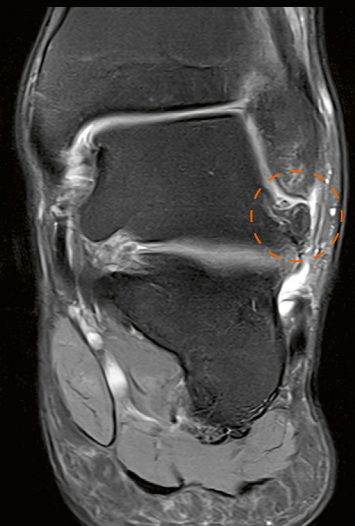
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MAGNETOM Flash — Imprint

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Publisher:**Siemens Healthineers AG**

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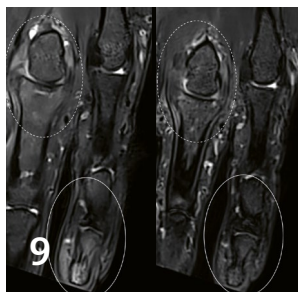
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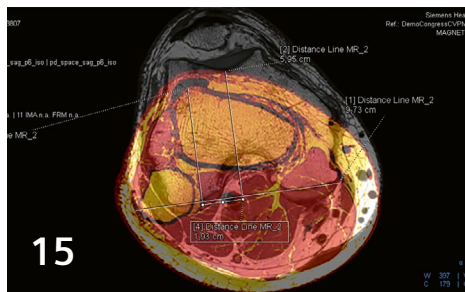
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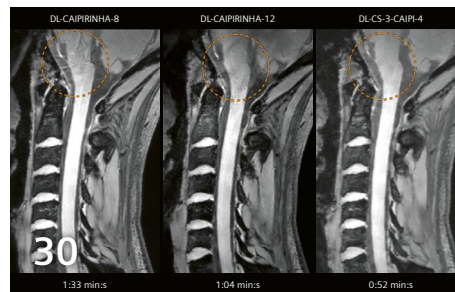
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One sequence,
multiple contrasts



How to Measure Tibial Tuberosity
to Trochlear Groove Distance



Highly Accelerated 7T MR Neurography

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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.

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From Better Images to Better Decisions: The Next Step for MSK MRI

Musculoskeletal MRI has never offered better images than it does today. Contemporary examinations combine high-field systems, optimized pulse sequences, three-dimensional acquisitions, and advanced acceleration and reconstruction. Structures that were once indistinct can now be displayed with remarkable clarity. Examinations that previously occupied long scanner slots can be completed within clinically practical times. Even in the presence of orthopedic hardware, diagnostic information can often be recovered from regions that were formerly obscured. This progress reflects years of collaboration between radiologists, physicists, engineers, technologists, and industry.

Yet better images are not the ultimate objective of medical imaging. A sharper cartilage surface, a more conspicuous nerve fascicle, or an image with less noise has value only if it helps us make a more accurate diagnosis, increases relevant diagnostic confidence, changes management, or improves patient experience or outcomes. Image quality is immediately visible. Clinical value is harder to demonstrate.

This year's International Skeletal Society (ISS) edition of MAGNETOM Flash highlights the extraordinary momentum of musculoskeletal MRI. It also provides an opportunity

to consider what should come next. I am convinced that the next step will not be defined by a single field strength, sequence, or artificial intelligence application, but rather by the transition from optimizing images to optimizing decisions. That transition requires us to ask not only whether a new technique performs better, but also what problem it solves, for whom, and with what consequences.

Image quality is the beginning, not the endpoint

Technical image quality remains essential. An examination affected by motion, insufficient resolution, inhomogeneous fat suppression, or severe susceptibility artifacts may conceal the relevant finding. Improvements in signal-to-noise ratio, contrast, sharpness, and artifact reduction therefore form the foundation of diagnostic progress. But that foundation is only the beginning.

The classic hierarchy proposed by Fryback and Thornbury remains strikingly relevant [1]. It begins with technical efficacy and then moves through diagnostic accuracy, influence on diagnostic thinking, influence on management, patient outcomes, and societal benefit. Success at a lower level is necessary, but it does not guarantee success at the level above. A sequence may receive excellent image-quality scores without improving

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lesion detection. Greater sensitivity may identify abnormalities whose natural history is uncertain. Additional findings may increase downstream testing without improving care.

For MSK MRI, this is not an abstract concern. We image many abnormalities that are common in asymptomatic individuals, including degenerative tendon changes, labral and meniscal variants, cartilage defects, and postoperative findings. As spatial resolution increases, so does the number of visible details. A high-resolution 7T MRI study that we conducted at Balgrist Campus, for example, demonstrated substantial variability in the meniscal root anatomy of the knee [2]. Such knowledge can deepen anatomical understanding and help prevent normal variants from being mistaken for tears. It also illustrates why seeing more must be accompanied by an understanding of what these additional findings mean.

Assessments of image quality and reader preference remain valuable, but a broader evaluation should also consider lesion detection, classification, interreader agreement, diagnostic confidence, and agreement with an appropriate reference standard. Whenever feasible, studies should also examine whether the information changes the radiologist's conclusion or the clinician's management plan. The most convincing innovation is not the one that produces the most impressive side-by-side images. It is the one that resolves a real diagnostic uncertainty.

Faster MRI: Time as a clinical resource

Acceleration is among the most visible developments in current MRI. Parallel imaging, simultaneous multislice acquisition, compressed sensing, and deep-learning reconstruction can substantially shorten acquisitions. Faster examinations can reduce discomfort, motion, and the likelihood that a patient cannot complete the study. They can facilitate imaging of children, patients in pain, and individuals who struggle to remain still. At a system level, shorter protocols improve access and allow scarce MRI capacity to serve more patients.

Nevertheless, speed should not become an isolated performance target. Reconstruction can alter image texture, suppress noise, and change the appearance of subtle structures. A fast sequence that looks smooth may still obscure a low-contrast lesion; conversely, an unfamiliar texture may be judged negatively despite preserved diagnostic information. Validation must therefore include the abnormalities and clinical decisions for which the sequence will be used, rather than relying solely on perceived quality in healthy volunteers.

Our work on deep-learning-reconstructed, high-resolution knee MRI at 7T illustrates both the opportunity and the necessary discipline [3]. Higher parallel-imaging acceleration could be achieved with useful image quality within feasible acquisition times, but performance depended on

the degree of acceleration. The broader lesson applies at every field strength: Acceleration is a trade-off, not free time. The dividend can be taken as a shorter examination, or reinvested in spatial resolution, three-dimensional coverage, more robust fat suppression, or an additional contrast. The correct choice depends on the patient and the clinical question.

For a routine follow-up examination, a shorter protocol may provide everything required. For suspected subtle cartilage injury, peripheral nerve pathology, or a postoperative complication, it may be better to invest the time saved in image detail. Protocol design should therefore become more adaptive. The goal is not the shortest possible MSK MRI, but rather the shortest examination that answers the specific question reliably.

When higher resolution changes the question

Ultra-high-field MRI makes the relationship between technical capability and clinical value especially clear. The increased signal available at 7T can be converted into exceptional spatial resolution. This has enabled visualization of structures such as cervical nerve rootlets, dorsal root ganglia, lumbar nerve fascicles, fine cartilage morphology, and small meniscal insertions. Such images expand our understanding of anatomy. However, their strongest justification arises when they change how disease is assessed.

In patients with cervical radiculopathy, high-resolution 7T double-echo steady-state (DESS) imaging enabled direct assessment of nerve-root compression and improved diagnostic confidence and prediction of sensory or motor dysfunction compared with standard clinical MRI [4]. Here, the advance is not simply that the nerve root appears sharper. This approach allows the radiologist to assess the affected neural structure more directly, rather than relying primarily on foraminal narrowing as an indirect sign of neural compromise. That represents a meaningful change in the diagnostic approach. In this issue, Abel et al. show that combining compressed sensing and CAIPIRINHA with deep-learning reconstruction can retain clear visualization of cervical nerve structures in highly accelerated 7T DESS neurography, even at acceleration factors where CAIPIRINHA alone becomes increasingly limited by artifacts [5]. This is a useful example of acceleration being used to make high spatial resolution more clinically practical.

Similarly, 7T MR neurography of the lumbosacral plexus improved visualization of nerve roots, dorsal root ganglia, and fascicles compared with acquisition-time-matched 3T imaging [6]. However, some artifacts were more pronounced at 7T. This balanced finding is instructive: Higher field strength does not improve every aspect of every examination, and technical superiority should never be assumed from field strength alone. The relevant

endpoint is whether the additional anatomical information is robust enough to improve diagnosis in the patients who need it.

From quantitative measurements to clinical value

Quantitative muscle MRI provides another example of how imaging has improved clinical decision-making. In patients with rotator cuff tears, two-point Dixon MRI enables volumetric measurement of muscle fat fraction, replacing a broad visual category with a continuous quantitative value. In one of our studies, preoperative fat fractions of the supraspinatus, infraspinatus, and subscapularis muscles provided muscle-specific cutoff values for predicting retear after arthroscopic repair [7]. These measurements transform fatty infiltration from a descriptive imaging finding into a clinically actionable risk marker that can inform patient selection, surgical planning, and counseling about the likelihood of durable tendon healing.

We applied a similar approach to chronic Achilles tendon disease, using two-point Dixon MRI to quantify calf-muscle fat content in patients with midportion or insertional disease [8]. The mean fat fraction of the triceps surae was approximately 11% and did not differ between the two disease locations, although their patterns of tendon damage and accompanying findings differed substantially. This result is also informative: Quantitative MRI can reveal meaningful differences, but it can also demonstrate when clinically distinct phenotypes share similar muscle composition. Such measurements add context to morphological findings and may provide a basis for future longitudinal and treatment-response studies.

Quantitative assessment is not limited to muscle. Ultrashort-echo-time (UTE) MRI extends this approach to tissues with short T2 relaxation times, including tendons, ligaments, and menisci, which are difficult to assess with conventional sequences. In a whole-knee 3T study, UTE T2* mapping demonstrated good scan–rescan repeatability, while also showing that knee position can influence measurements in some tissues [9]. UTE magnetization-transfer approaches offer measurements of tendon degeneration that are less sensitive to orientation effects than UTE T2* mapping [10]. Together, these studies show that technical capability is only one part of the path toward clinical value; quantitative measurements must also be reproducible, biologically interpretable, and relevant to diagnosis or patient management.

Restoring diagnostic value after surgery

Postoperative imaging offers perhaps the clearest example of technology solving an unmet clinical need. Orthopedic metal implants can cause signal loss, distortion, pile-up, and failure of fat suppression on MRI. Without appropriate

techniques, the tissues closest to the implant may be precisely those we cannot evaluate. Basic MRI protocol choices, including field strength, bandwidth, voxel size, pulse sequence, and fat suppression, remain important. Advanced methods such as compressed-sensing slice encoding for metal-artifact correction (CS-SEMAC) can further recover anatomy near hardware [11].

The clinical benefit is substantial because MRI can evaluate the implant environment and the surrounding soft tissues in one examination. Metal-artifact-reduced MRI has demonstrated diagnostic value in challenging settings such as suspected periprosthetic hip infection [12]. It can also depict synovitis, osteolysis, muscle injury, tendon failure, collections, and neurovascular abnormalities. In this context, artifact reduction does more than improve appearance; it gives the radiologist access to a previously diagnostically inaccessible region and may influence the decision between observation, aspiration, revision surgery, or another intervention.

Even here, however, better images require informed interpretation. Metal-artifact-reduction techniques can introduce characteristic blurring and ripple artifacts, although newer approaches have shown considerable success in reducing these artifacts [13]. The remaining artifact pattern is influenced by implant material and geometry, while interpretation also requires knowledge of the surgical approach and expected postoperative changes. Technology opens the diagnostic window; expertise determines what we see through it.

From validation to implementation

Moving from better images to better decisions requires a broader culture of validation. First, we need technical studies that define how a method behaves, including its failure modes. Second, diagnostic studies should use realistic patient populations, relevant pathology, multiple readers, and appropriate reference standards. Third, workflow studies should determine whether the method changes reporting time, confidence, recommendations, or communication with referring clinicians. Finally, outcome and health-system studies should examine whether the technique improves care, avoids unnecessary tests, reduces repeat imaging, or expands access.

Not every innovation needs a randomized trial before clinical use. The level of evidence should be proportional to the claim being made. If a reconstruction is proposed as a faster replacement for an established sequence, diagnostic noninferiority may be the central requirement. If a quantitative biomarker is claimed to guide therapy, longitudinal and outcome data become essential. If an automated tool influences interpretation, its performance must be assessed across institutions, scanners, patient groups, and uncommon but consequential findings.

Implementation also requires attention to generalizability. Many advances originate in highly specialized centers with expert technologists, optimized coils, and dedicated physicists. Their broader value depends on reproducibility in routine practice. A method that performs exceptionally under ideal conditions but fails unpredictably during a busy clinical day will not improve decisions. Robustness, ease of use, and education are therefore clinical performance characteristics, not secondary operational details.

The next step

As new MSK MRI techniques emerge, three questions can keep innovation clinically grounded. First, does the technique reveal relevant information that was previously unavailable or unreliable? Second, does it provide the necessary information faster, more safely, or with less burden on the patient? Third, does the result change the diagnostic thinking, management, or outcome? A technique need not satisfy all three, but its intended contribution should be explicit.

This framework also helps us choose wisely among competing possibilities. Sometimes the best decision will be to use ultra-high resolution; at other times it will be to accept less detail in exchange for speed and comfort. Sometimes a quantitative sequence will add important biological information; at other times a well-optimized conventional protocol will be the correct examination. Innovation should expand our options, not make every protocol more complicated.

This edition of MAGNETOM Flash accompanies the Annual Meeting of the International Skeletal Society. The advances presented in this ISS edition reflect how dynamic MSK MRI has become. Across the field, we can acquire images faster, reconstruct them more intelligently, visualize smaller structures, quantify tissue properties, and obtain diagnostic information closer to metal than was possible only a few years ago. The responsibility that accompanies these capabilities is to connect them to the decisions faced by patients and clinicians.

Better images will always inspire us. They reveal anatomy, stimulate new questions, and drive the field forward. But the enduring measure of progress is not how an image performs on the screen; it is whether the information helps the right patient receive the right diagnosis and the right care. The next step for musculoskeletal MRI is therefore clear: from better images to better decisions.



Reto Sutter

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One Sequence, Multiple Contrasts: Toward a Simplified Knee MRI Protocol with 3D PD TSE nDixon

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1. Introduction

Magnetic resonance imaging (MRI) of the knee remains one of the most frequently performed musculoskeletal examinations worldwide. Despite major advances in acquisition acceleration, routine protocols still rely on multiple two-dimensional (2D) sequences acquired in different planes and contrasts. A typical examination combines several proton density (PD) fat-suppressed sequences with a dedicated T1-weighted acquisition for bone marrow assessment, resulting in scan times that frequently exceed 10 minutes.

Reducing examination time while maintaining diagnostic confidence has become a major objective in musculoskeletal MRI. Beyond improving patient comfort, shorter protocols increase scanner throughput, reduce the likelihood of motion-related image degradation, and improve access to MRI.

Recent developments in isotropic three-dimensional turbo spin-echo (3D TSE) imaging have substantially changed this landscape. Modern volumetric acquisitions now provide image contrast comparable to conventional 2D TSE imaging while enabling high-quality multiplanar reformations from a single dataset. Combined with advances in deep-learning reconstruction, image quality

and spatial resolution continue to improve while acquisition times decrease [1].

At the same time, Dixon fat–water separation has emerged as an attractive alternative to conventional spectral fat suppression. In addition to providing homogeneous fat suppression, Dixon reconstruction simultaneously generates water-only, fat-only, in-phase, and opposed-phase images from a single acquisition, thereby expanding the available diagnostic information without prolonging scan time [2].

Over the past decade, Siemens Healthineers has promoted this evolution through concepts such as GOKnee3D, demonstrating that comprehensive knee assessment can be achieved using a limited number of isotropic 3D acquisitions. The next logical step is to move beyond replacing individual 2D sequences with several volumetric acquisitions and instead perform a complete routine knee MRI examination using a single isotropic dataset [3].

The 3D PD TSE nDixon sequence¹ represents this new approach by combining isotropic 3D imaging, Dixon fat–water separation, and deep-learning reconstruction into a single comprehensive acquisition.

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 concept: Combining 3D imaging and Dixon reconstruction

The 3D PD TSE nDixon sequence is acquired in less than six minutes and combines three complementary technologies: isotropic 3D proton-density turbo spin-echo imaging, Dixon fat–water separation, and deep-learning based image reconstruction.

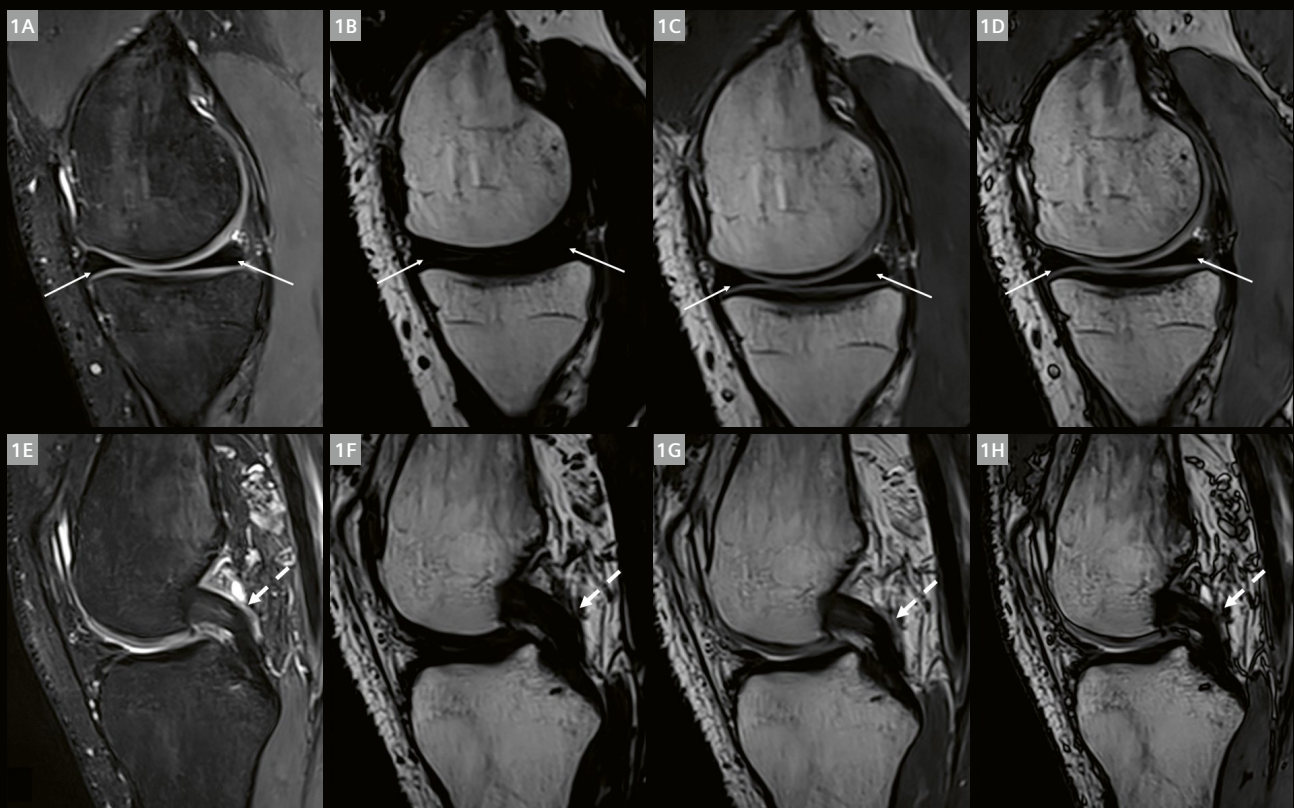
Unlike conventional 3D fat-suppressed TSE sequences that rely on spectral attenuated inversion recovery (SPAIR), Dixon-based reconstruction performs intrinsic fat–water separation and is considerably less sensitive to B0 field inhomogeneities. This advantage is particularly relevant in peripheral musculoskeletal imaging, where complex anatomy and off-center positioning frequently compromise conventional fat suppression.

From a single acquisition, Dixon reconstruction simultaneously provides water-only images equivalent to conventional PD fat-suppressed imaging together with fat-only, in-phase, and opposed-phase reconstructions.

Multiple clinically useful image contrasts are therefore generated without increasing acquisition time.

The availability of fat-only images is particularly valuable because they reproduce much of the information traditionally obtained from a dedicated T1-weighted sequence. Bone marrow composition, fatty replacement, marrow reconversion, incidental marrow lesions, fractures, and many bone abnormalities can therefore be evaluated without requiring an additional acquisition [4]. Similar concepts have already been successfully implemented in other musculoskeletal applications, including sacroiliac joint imaging [5].

Together, isotropic imaging and Dixon reconstruction create the possibility of performing a comprehensive routine knee examination using a single volumetric acquisition (Fig. 1).



1 Representative normal knee MRI obtained with the 3D PD TSE nDixon sequence.¹ Images demonstrate the anterior and posterior horns of the medial meniscus (solid white arrows; 1A–1D) and the posterior cruciate ligament (dotted white arrow; 1E–1H). (1A, 1E) Water-only images; (1B, 1F) fat-only images; (1C, 1G) in-phase images; and (1D, 1H) opposed-phase images.

3. Clinical validation

We prospectively evaluated this concept by comparing a single 3D PD TSE nDixon acquisition with a conventional high-resolution knee MRI protocol consisting of four separate 2D TSE sequences, including a dedicated T1-weighted acquisition [3].

The conventional protocol required 12 minutes 6 seconds, whereas the complete 3D PD TSE nDixon examination required only 5 minutes 55 seconds, corresponding to an approximately 50% reduction in acquisition time.

Despite this substantial reduction, diagnostic performance remained comparable for the abnormalities most frequently encountered in daily practice, including meniscal tears, anterior cruciate ligament injuries (Fig. 2), collateral ligament injuries, fractures (Fig. 3), and bone marrow edema. No significant differences in diagnostic accuracy were observed between the two approaches.

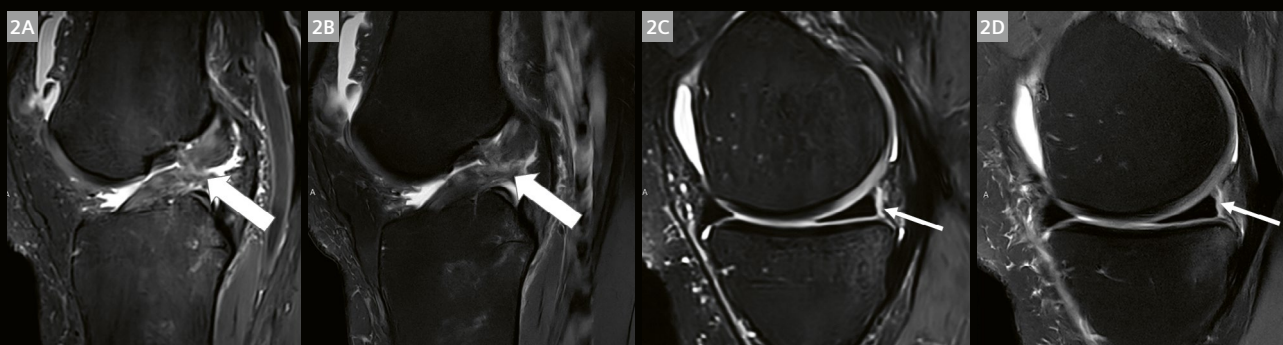
The isotropic nature of the dataset also enabled high-quality multiplanar reformations tailored to the anatomy of interest, facilitating the evaluation of complex lesions such

as radial meniscal tears and ramp lesions while reducing partial-volume effects (Fig. 4). This also resulted in better interobserver agreement.

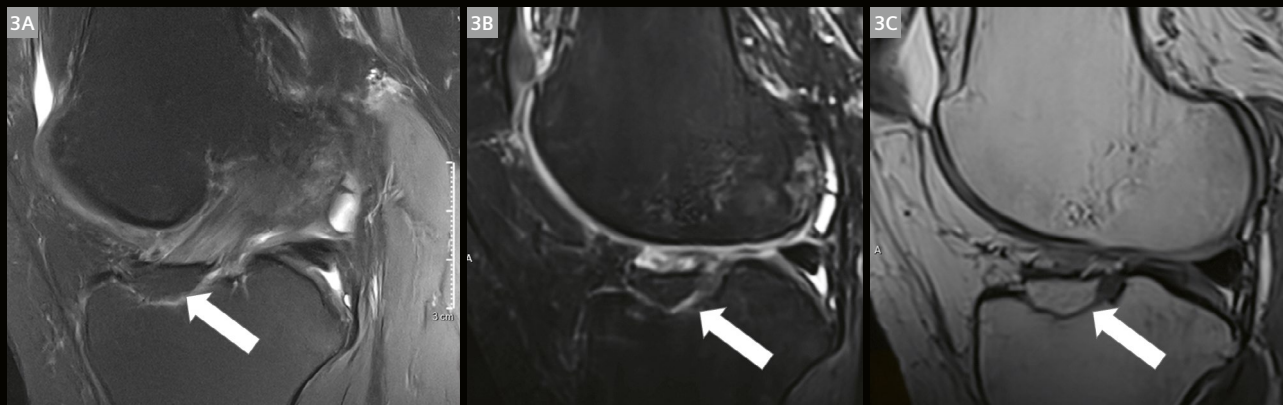
4. Why Dixon changes the workflow

Beyond improving fat suppression, Dixon fundamentally changes the way routine musculoskeletal MRI examinations are interpreted: First, fat suppression is generally more homogeneous and less sensitive to B0 field inhomogeneities than conventional spectral fat-saturation techniques. Second, water-only and fat-only images are generated simultaneously from the same acquisition. In routine knee MRI, radiologists no longer need to alternate between PD fat-suppressed and T1-weighted sequences to assess bone marrow. Both types of information are immediately available within the same volumetric dataset.

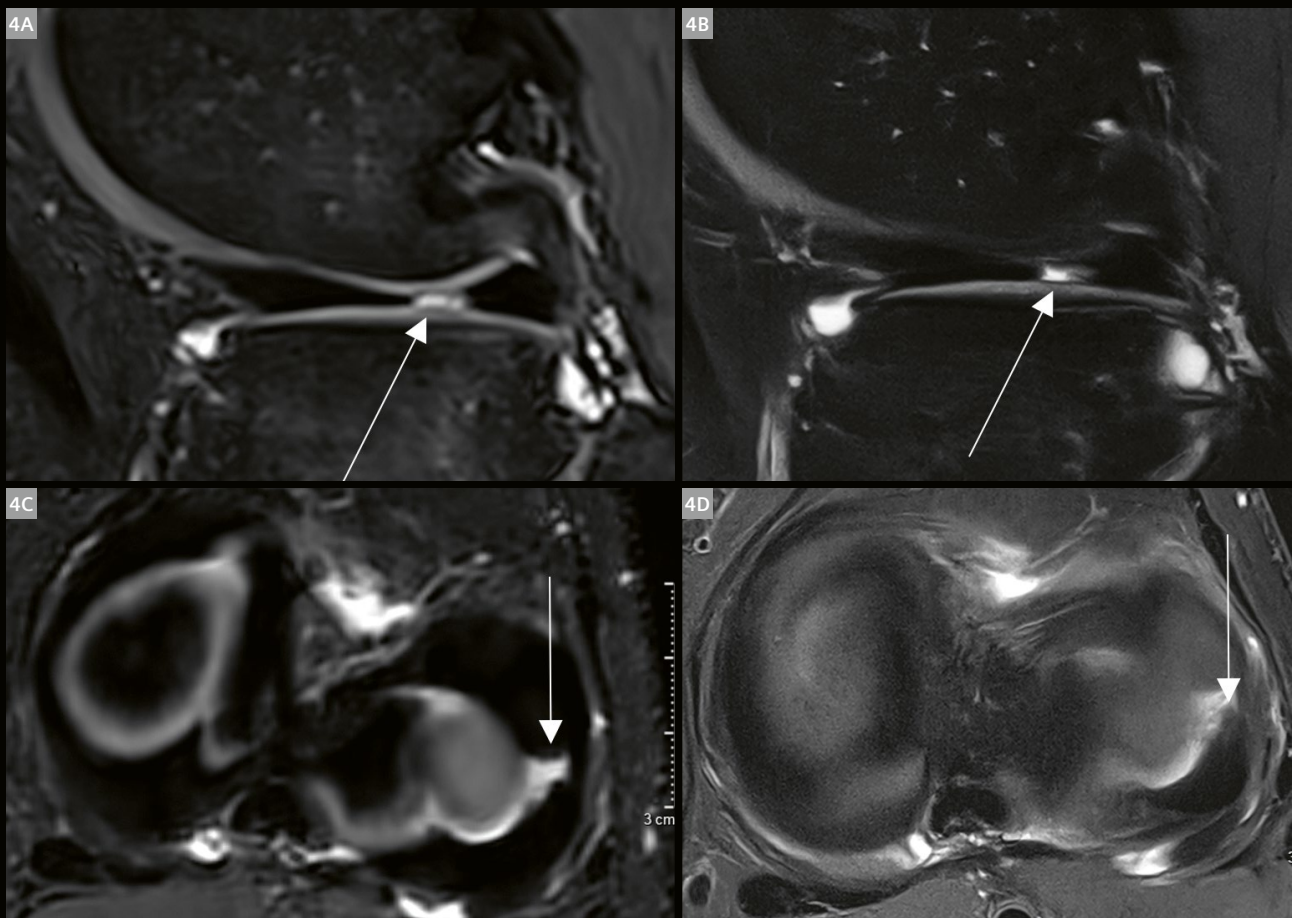
This simplified workflow is particularly valuable when evaluating bone marrow edema, stress injuries, marrow reconversion, osteonecrosis, incidental marrow abnormalities, and focal bone lesions (Fig. 5). Beyond the knee,



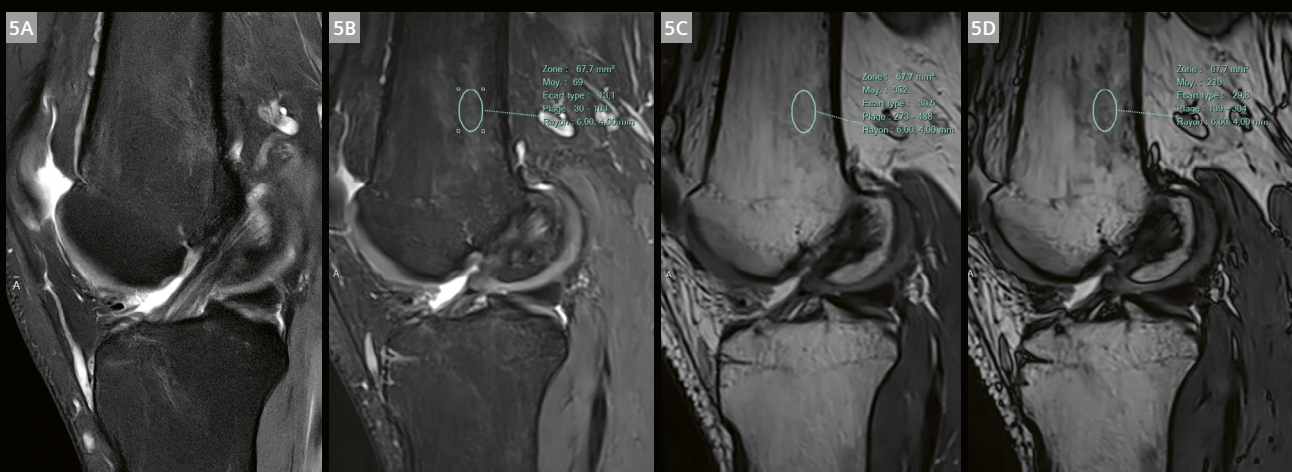
2 Comparison between 3D PD TSE nDixon and conventional MRI. Anterior cruciate ligament tear (thick white arrow; 2A, 2B) and ramp lesion of the medial meniscus (white arrow; 2C, 2D). (2A, 2C) Water-only images from the 3D PD TSE nDixon acquisition; (2B, 2D) corresponding conventional high-resolution fat-suppressed PD TSE images.



3 Comparison between 3D PD TSE nDixon and conventional MRI in a patient with an anterior cruciate ligament enthesis avulsion fracture (thick white arrow). (3A) Conventional high-resolution fat-suppressed PD TSE image. (3B) Water-only image from the 3D PD TSE nDixon acquisition. (3C) Fat-only image from the 3D PD TSE nDixon acquisition.



4 Comparison between conventional MRI and 3D PD TSE nDixon in a patient with a radial tear of the lateral meniscus (white arrow). (4A, 4C) Water-only images from the 3D PD TSE nDixon acquisition; (4B, 4D) corresponding conventional high-resolution fat-suppressed PD TSE images.



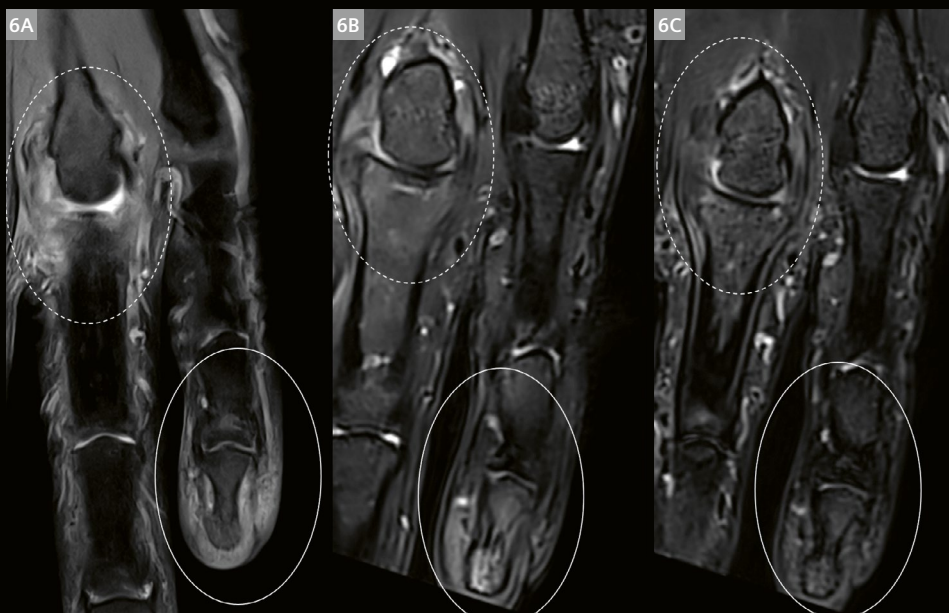
5 Benign red marrow hyperplasia. The lesion is visible on the conventional high-resolution fat-suppressed PD TSE image (5A) and on the water-only image from the 3D PD TSE nDixon acquisition (5B). The in-phase (5C) and opposed-phase (5D) reconstructions demonstrate signal loss on the opposed-phase image, consistent with benign marrow hyperplasia.

similar advantages have already been demonstrated in other anatomical regions [4–7].

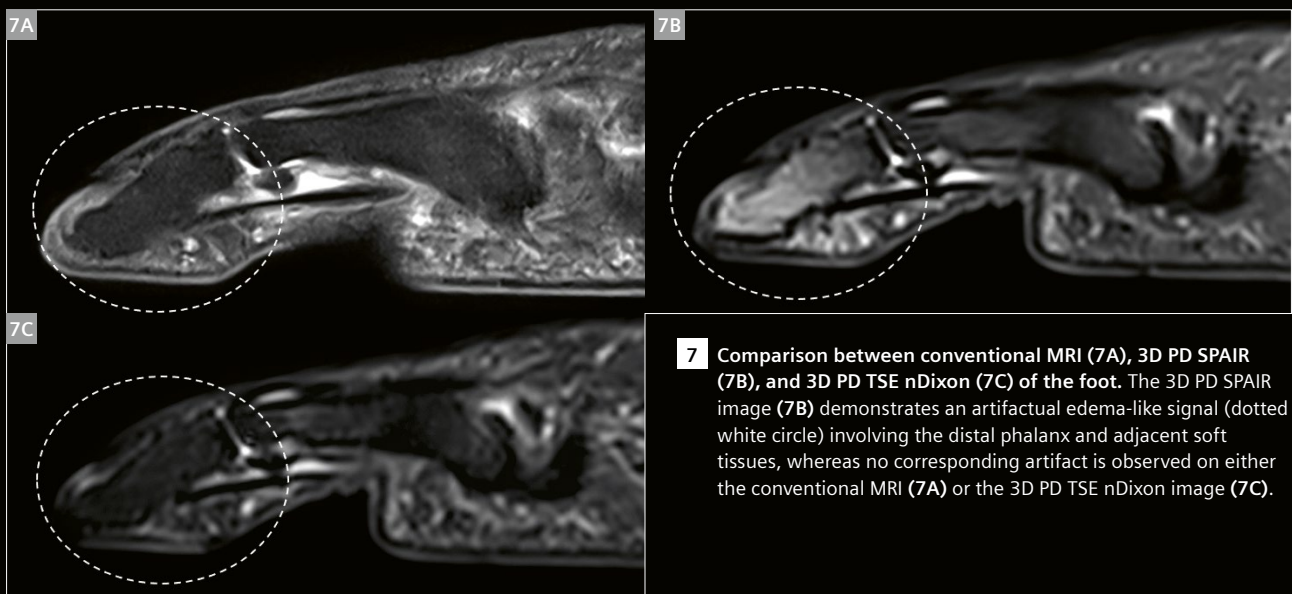
5. Motion robustness: An additional practical advantage

Although motion robustness is not an intrinsic property of the sequence itself, replacing multiple conventional acquisitions with a single volumetric dataset substantially reduces opportunities for patient movement during the examination.

In everyday practice, motion artifacts frequently occur toward the end of MRI examinations, particularly in patients with acute trauma or significant pain. By shortening the overall examination and eliminating repeated repositioning between sequences, the single-acquisition approach naturally decreases the likelihood of motion-related image degradation. This benefit may be particularly relevant for pediatric patients, elderly individuals, and patients who have difficulty remaining still during prolonged examinations.



6 Comparison between conventional MRI (6A), 3D PD SPAIR (6B), and 3D PD TSE nDixon (6C) in a patient with a traumatic capsular injury of the fourth metacarpophalangeal joint (white arrow). The capsular lesion is well depicted on all three sequences. However, spurious edema-like signal (white circle) involving the distal phalanx and adjacent soft tissues is present on the 3D PD SPAIR image (6B), while a similar artifact is visible in the surrounding soft tissues on the conventional MRI (6A). No such abnormality is observed on the 3D PD TSE nDixon image (6C).



7 Comparison between conventional MRI (7A), 3D PD SPAIR (7B), and 3D PD TSE nDixon (7C) of the foot. The 3D PD SPAIR image (7B) demonstrates an artifactual edema-like signal (dotted white circle) involving the distal phalanx and adjacent soft tissues, whereas no corresponding artifact is observed on either the conventional MRI (7A) or the 3D PD TSE nDixon image (7C).

6. Beyond the knee: Extending the concept to small joints

Although the knee is an ideal proof of concept, the potential of 3D PD TSE nDixon imaging extends well beyond this application. Small joints of the wrist, hand, fingers, ankle, and foot remain challenging because of their complex anatomy and the susceptibility of conventional SPAIR-based fat suppression to peripheral magnetic field inhomogeneities (Fig. 6). In these anatomical regions, isotropic 3D imaging allows multiplanar reformations precisely aligned with small ligaments, tendon insertions, pulleys, cartilage surfaces, and osteochondral structures, while Dixon reconstruction provides more robust fat suppression throughout the field of view (Fig. 7). Our initial experience suggests that these applications may benefit even more from the combination of isotropic imaging and Dixon reconstruction than the knee itself, and investigations are currently evaluating these indications.

7. Looking ahead

For many years, musculoskeletal MRI has sought to reduce the number of acquisitions without compromising diagnostic performance. The combination of isotropic 3D TSE imaging, Dixon fat–water separation, and deep-learning reconstruction brings this objective closer to routine clinical practice. Rather than considering 3D imaging as an additional sequence, it may now be viewed as the foundation of a comprehensive examination.

Our clinical experience demonstrates that a single 3D PD TSE nDixon acquisition provides comprehensive evaluation of menisci, ligaments, cartilage, fractures, and bone marrow abnormalities while reducing acquisition time by approximately 50%. The simultaneous availability of water-only and fat-only reconstructions further simplifies the examination by reducing dependence on dedicated T1-weighted imaging.

Rather than representing another incremental technical improvement, single-acquisition isotropic Dixon imaging may mark a genuine shift in musculoskeletal MRI — from protocols built around multiple individual sequences to comprehensive volumetric imaging capable of delivering the entire examination from a single acquisition.

Key benefits of 3D PD TSE nDixon

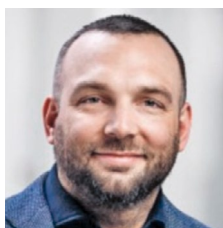
- One acquisition, four image contrasts
- Potential replacement of dedicated T1 imaging
- ~ 50% shorter examination time
- Isotropic multiplanar reformations
- Robust Dixon fat suppression
- Deep-learning reconstruction
- Motion-resistant workflow
- Potential extension to hand, wrist, ankle, and foot imaging

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How to Measure the Tibial Tuberosity to Trochlear Groove Distance on MRI

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The tibial tuberosity to trochlear groove (TT-TG) distance is a radiographic measurement used primarily to assess patellar instability (which occurs when the kneecap slips out of place). It measures the lateral offset between the deepest part of the trochlear groove and the point where the patellar tendon attaches to the shinbone.

How often is this measurement requested?

The TT-TG measurement is often requested in orthopedic clinics, specifically for patients presenting with the following issues:

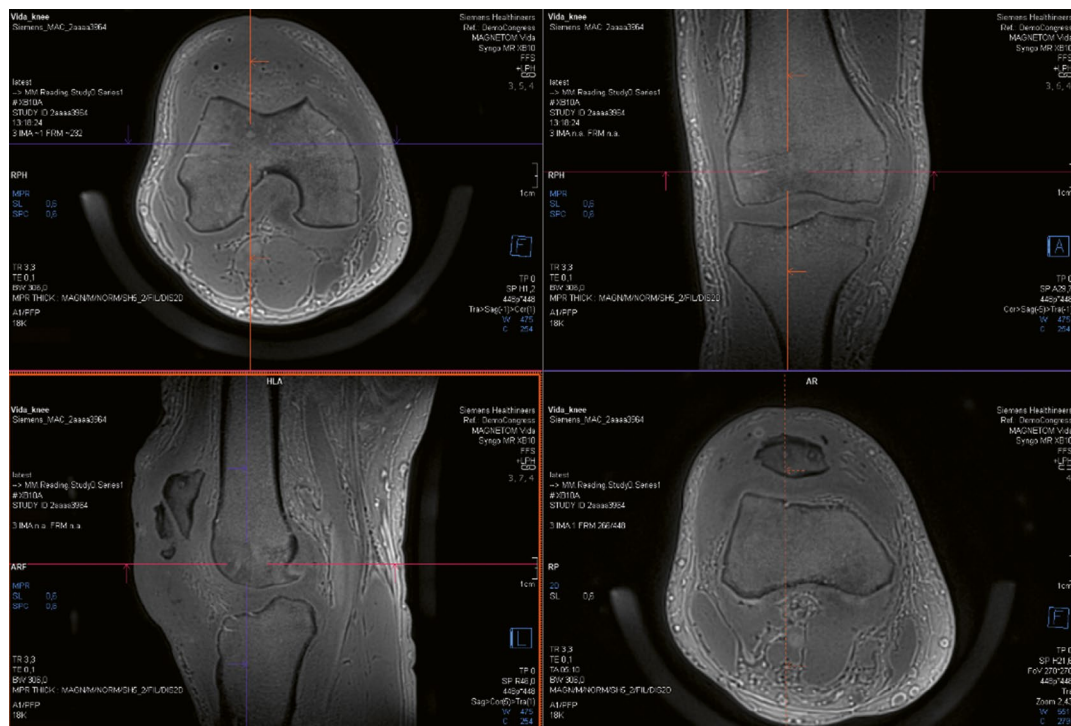
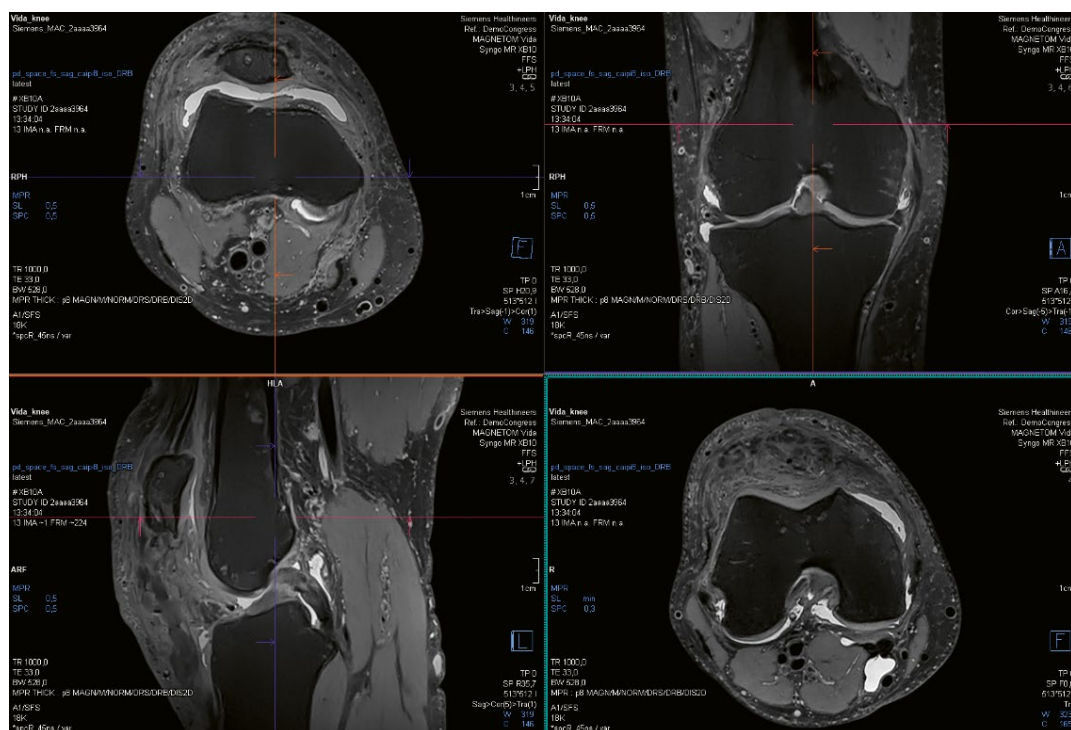
- Recurrent patellar dislocations
- Chronic kneecap pain (patellofemoral pain syndrome)
- Patellar maltracking

If a surgeon is considering an operation to stabilize the kneecap, the TT-TG distance is part of the standard preoperative workup.

The measurement is typically performed using computed tomography (CT) or magnetic resonance imaging (MRI). Radiologists take two cross-sectional images: one at the level of the trochlea and one at the level of the tibial tuberosity. They then superimpose the images to measure the horizontal distance between the two points.

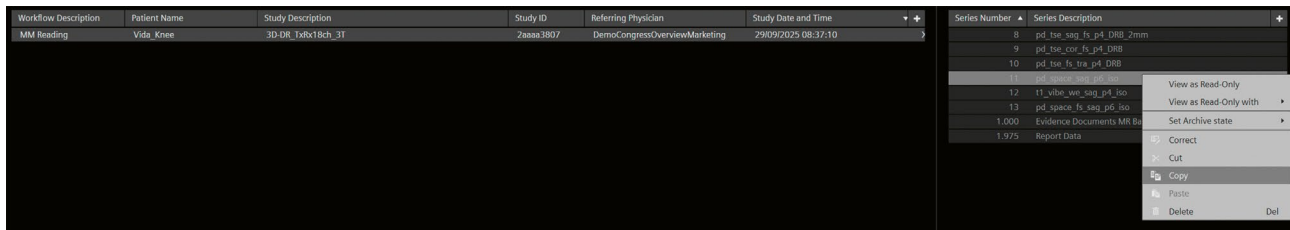
Historically, CT was the gold standard examination for this measurement, partially because the modality can provide 3D acquisitions. Nowadays, MRI can perform 3D acquisitions with thin slices very similar to CT, and it brings additional advantages such as different contrasts (achieved with T1, T2, PD, fat-sat, or zero echo time imaging PETRA) that show the cartilage and ligaments at the same time without ionizing radiation. Another advantage of MRI is the possibility of reliably measuring the tibial tuberosity to posterior cruciate ligament distance (TT-PCL). This information helps determine if the instability failure is purely due to the tibial bone.

The acquisition time for an isotropic 3D T1 SPACE sequence with 0.33 mm slice thickness is around 3 minutes. If the radiologist requests a 3D FS acquisition, it can be done in around 4 minutes (Fig. 1). Even the PETRA sequence can be done with 0.6 mm slice thickness (Fig. 2).

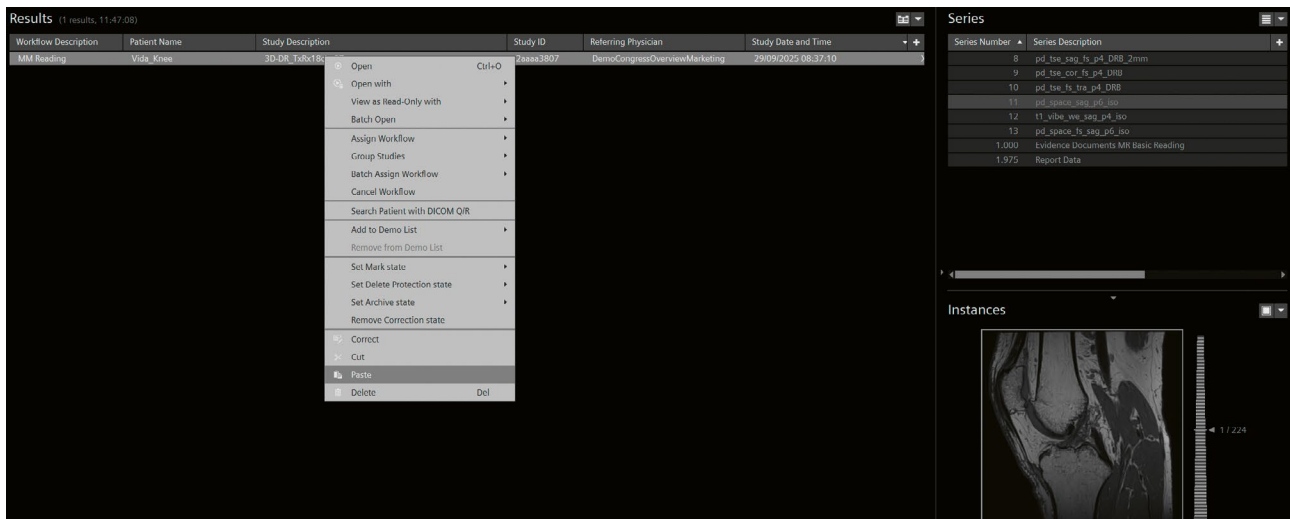


Here is a step-by-step guide for measuring the TT-GT distance.

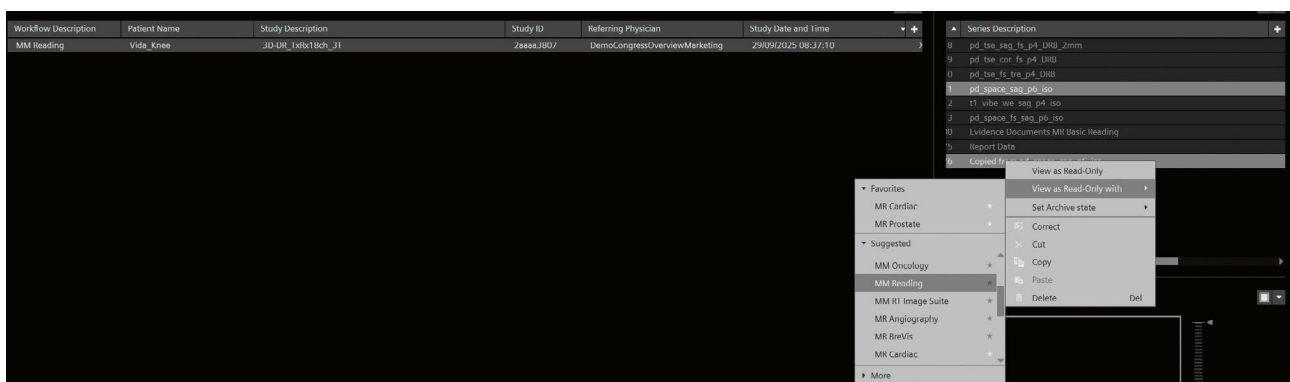
1) Choose the 3D acquisition and copy the sequence from the Series level.



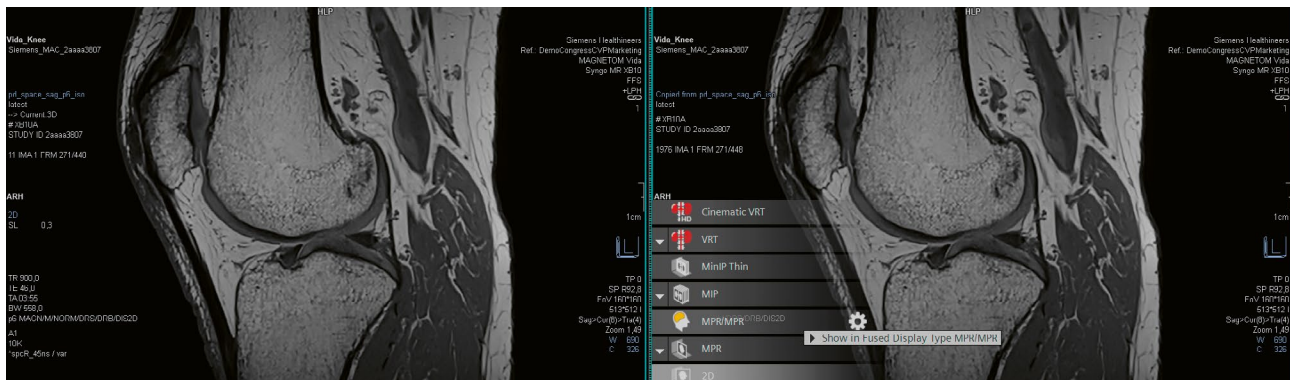
2) Select the patient's name and paste the sequence on the Results level.



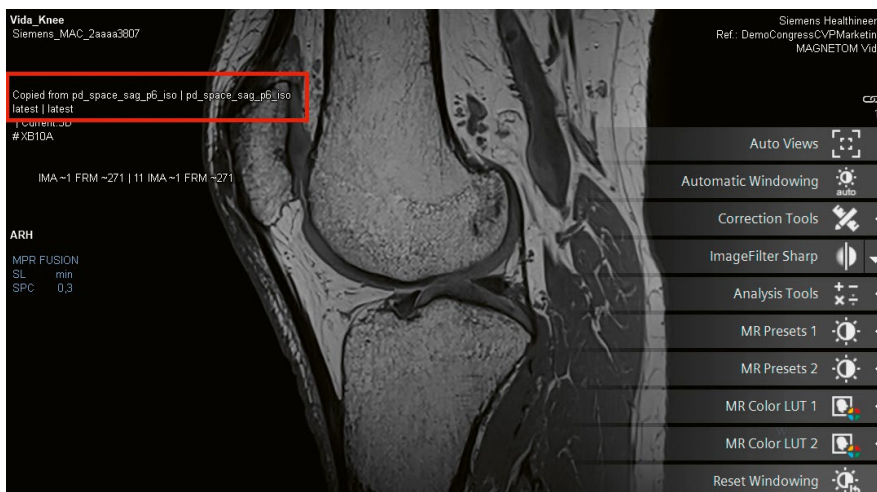
3) Select both series on the Series level and open them on the syngo.via multimodality reading workflow (MM Reading).



- 4) Press and hold the Ctrl key to select both images. Use the MPR/MPR option in the bottom left corner to fuse the images.



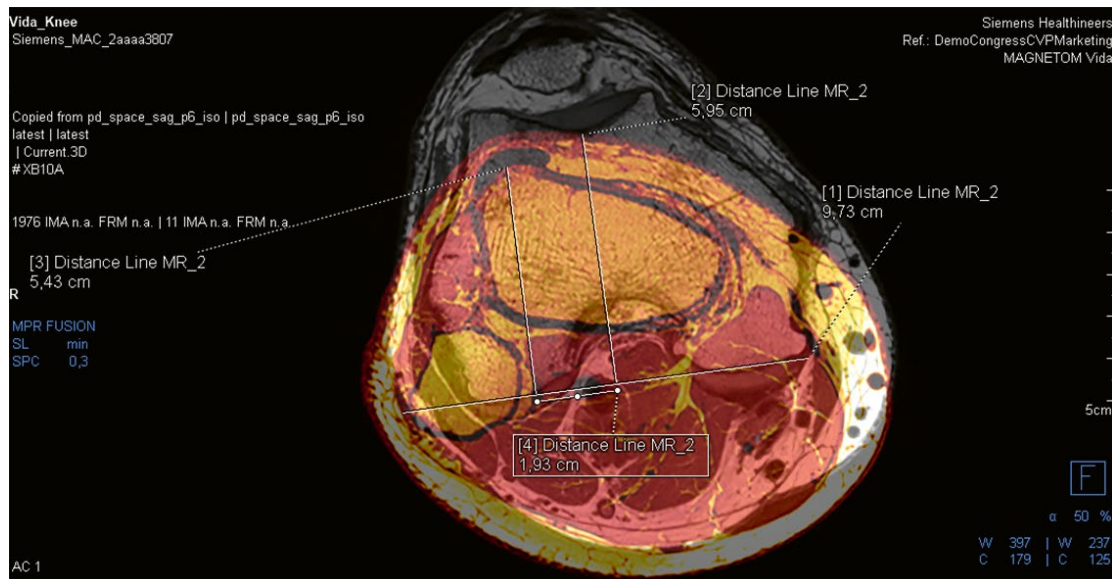
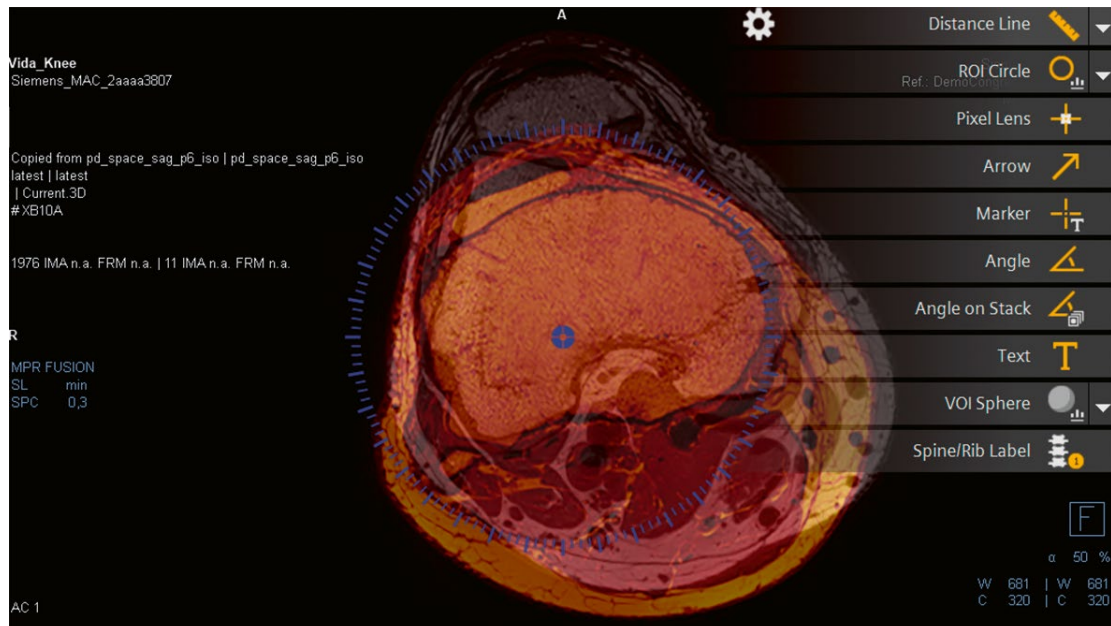
- 5) Define a color lookup table (LUT) for one of the images (MR Color LUT 2).
You can easily confirm the fusion sequence by checking the sequence names (they should be a combination of the two names).



- 6) Change the orientation to transversal view and open the Visual Alignment option in the top left corner.

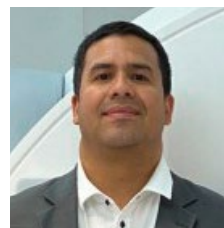


- 7) Press the Ctrl and Shift keys to scroll the images in one volume. In the top right corner, select Distance Line or Angle to do the measuring.



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Integrated Three-Dimensional MR Neurography: Innovations for Plexus and Peripheral Nerve Imaging

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Introduction

Magnetic resonance neurography (MRN) has emerged as an indispensable imaging technique for comprehensive evaluation of plexus and peripheral nerve disorders [1, 2]. The clinical utility of MRN depends critically on three fundamental technical requirements: (1) high spatial resolution, (2) homogeneous fat suppression, and (3) effective vascular suppression.

This article reviews integrated 3D MR neurography techniques and addresses technical challenges. The SPACE sequence incorporates advanced acceleration (DL CS CAIPI¹), fat suppression (C-FOCI, f-Dixon¹), and vascular suppression (MSDE¹, DANTE¹). Additionally, DESS with deep learning reconstruction¹ provides complementary imaging capabilities. Here we present practical protocol implementations and clinical applications that optimize imaging quality for plexus and peripheral nerve evaluation.

Integrated 3D SPACE: Combined acceleration, fat suppression, and vascular suppression

(1) Spatial resolution requirements for 3D MR neurography

High-resolution imaging is essential because peripheral nerves are thin, elongated structures composed of multiple fascicles, and pathologic changes may be subtle or focal. High-resolution 3-Tesla MRN substantially enhances nerve visualization and directly influences diagnostic accuracy and subsequent therapeutic decisions, making improved spatial resolution not merely a technical advancement but a clinically significant improvement in patient care [3]. Achieving high-resolution images of peripheral nerves

requires extended acquisition times. However, recent technological advances in the 3D SPACE sequence address this limitation through an integrated combination of three complementary techniques: parallel imaging with controlled aliasing (CAIPI), compressed sensing (CS), and deep learning (DL) reconstruction.

Parallel imaging with controlled aliasing (CAIPI)

Parallel imaging (PI) exploits differences in receiver coil sensitivity to reconstruct images from undersampled k -space data, reducing acquisition time. However, as acceleration increases, so does the g -factor penalty — noise amplification arising from the geometry and sensitivity profiles of the receiver coil array. CAIPIRINHA (Controlled Aliasing In Parallel Imaging Results In Higher Acceleration), also known as CAIPI, strategically shifts the phase-encoding pattern across slices in 3D acquisitions, reorganizing the aliasing artifacts that result from undersampling into regular, predictable patterns favorable for GRAPPA-based reconstruction [4]. This controlled aliasing pattern reduces g -factor penalties and enables higher acceleration factors while preserving signal-to-noise ratio.

Incoherent sampling and compressed sensing

Compressed sensing (CS) uses the inherent sparsity of medical images — the principle that images contain redundant information. Unlike uniform k -space undersampling, CS uses variable-density, pseudorandom patterns: Low frequencies are densely sampled to preserve contrast and overall image structure, while high-frequency data are undersampled irregularly [5]. This pseudorandom undersampling pattern produces incoherent, noise-like aliasing artifacts rather than coherent folding artifacts. Iterative reconstruction algorithms can then suppress these noise-like artifacts more effectively under data-consistency constraints, which enforce agreement between the reconstructed image and measured k -space

¹ Work 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.

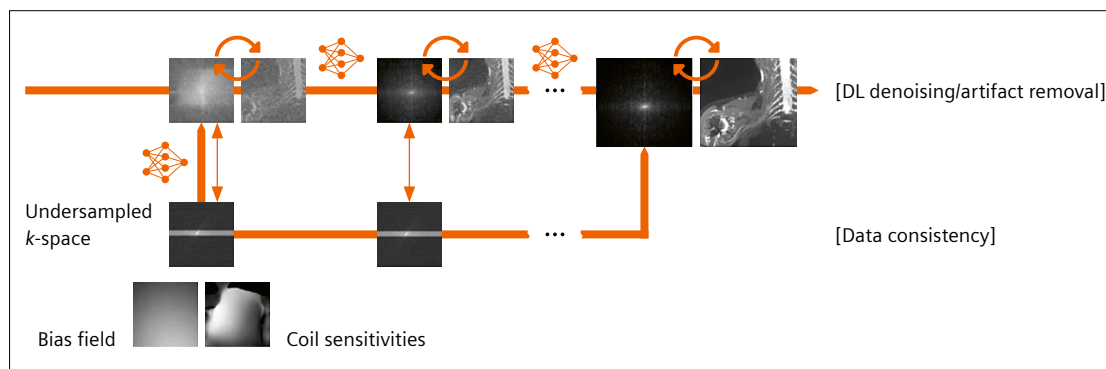
data. However, when pushed to very high acceleration factors, conventional iterative CS may introduce image smoothing — a critical limitation in neurography, where subtle fascicular details and focal signal abnormalities are diagnostically important.

Deep learning reconstruction

Deep Resolve Boost employs learned iterative networks to improve reconstruction. Rather than applying hand-crafted mathematical rules like in traditional reconstructions, the variational network framework implements iterative reconstruction as stacked neural network layers [6]. Each layer in the network alternates between enforcing data consistency in k -space and applying learned image refinements that suppress artifacts while preserving fine anatomic detail (Fig. 1). This iterative refinement through learned layers allows the network to progressively improve the reconstructed image over multiple passes, adapting to

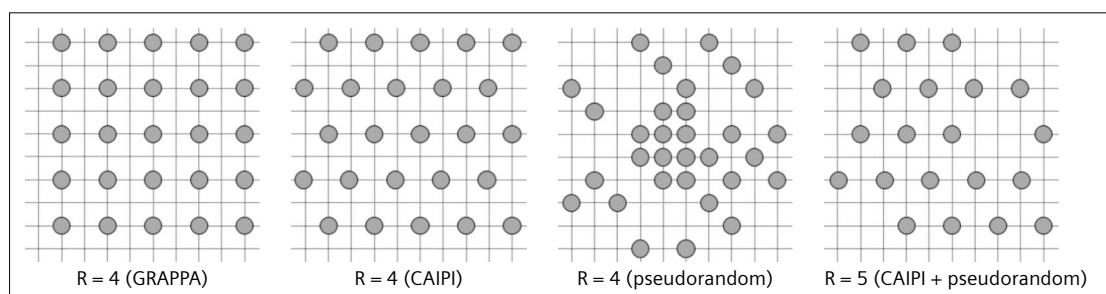
the specific undersampling pattern. Because the acquisition still relies on parallel imaging, performance ultimately remains constrained by coil geometry and g -factor penalties.

In contrast, CS-based DL reconstruction extends the acceleration strategy beyond conventional parallel imaging by incorporating incoherent undersampling and learned regularization into the reconstruction process. CS-based DL reconstruction learns a more flexible image prior from data, enabling more effective suppression of noise and undersampling artifacts while better preserving subtle anatomic structures. DL CS CAIPI reconstruction further combines the complementary strengths of parallel imaging, compressed sensing, and deep learning [7]. It allows imaging to move into higher acceleration ranges where aliasing is expected, but it designs the sampling pattern so that the aliasing becomes easier for the DL reconstruction algorithm to handle (Fig. 2). The result is high acceleration ($R \geq 5$) with preserved fine structural detail — essential



1 Schematic diagram of Deep Resolve.

Similar to conventional iterative compressed sensing, which alternates between data consistency (k -space term) and image regularization (denoising term), the variational network mimics this structure through stacked convolutional neural network (CNN) layers. Each layer enforces data consistency in k -space and applies learned CNN-based image refinement — replacing hand-crafted denoising with learned denoising from training data. Through multiple iterations, the network progressively reconstructs high-quality images while maintaining fidelity to measured k -space data.



2 k -space sampling patterns for different acceleration strategies.

$R = 4$ GRAPPA shows uniform, regular sampling that is favorable for parallel imaging reconstruction with predictable aliasing. $R = 4$ CAIPI demonstrates controlled phase-encoding shifts across slices that reorganize aliasing patterns while maintaining GRAPPA reconstruction efficiency. $R = 4$ pseudorandom sampling introduces variable-density, incoherent undersampling that transforms aliasing into noise-like artifacts that are suitable for compressed sensing and deep learning suppression. $R = 5$ CAIPI + pseudorandom (hybrid) combines the structured CAIPI pattern with pseudorandom variable-density undersampling, enabling higher acceleration while preserving the benefits of both parallel imaging and incoherent artifact suppression. The gray circles represent sampled k -space points.

for resolving small nerve fascicles and detecting focal pathology. These advances enable high-resolution 3D nerve imaging within clinically tolerable scan times, allowing better detection of subtle focal abnormalities and superior coverage of complex plexal structures.

Clinical implications for 3D MR neurography

MR neurography requires simultaneous achievement of high spatial resolution, adequate signal-to-noise ratio, and large field-of-view coverage. Conventional imaging approaches face inherent limitations in meeting all three requirements within acceptable scan times. The integrated DL CS CAIPI approach overcomes these constraints by enabling high acceleration ($R \geq 5$) while preserving image quality, allowing high-resolution 3D imaging within clinically tolerable scan times. Representative imaging protocols with acquisition parameters and expected scan times are summarized in tables 1–3. The approach

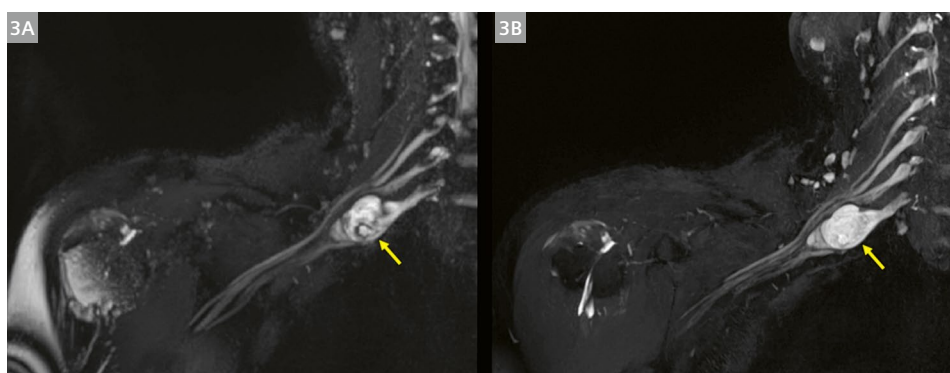
directly translates to superior visualization of nerve anatomy. This enables confident detection of pathologic changes, from nerve compressions and signal abnormalities to mass lesions (Fig. 3), while the ability to resolve individual nerve fascicles and small branch nerves enhances diagnostic accuracy.

(2) Fat suppression challenges

Uniform fat suppression is fundamental in MR neurography. Incomplete fat suppression obscures abnormal T2 hyperintensity and compromises nerve-to-background contrast, frequently resulting in diagnostic errors [8]. Fat suppression techniques used in MR neurography can be broadly categorized into three groups: chemical shift selective fat saturation (CHESS), inversion recovery techniques such as short inversion time inversion recovery (STIR), and chemical shift-based Dixon techniques.

Sequence	STIR SPACE	Deep Resolve STIR DANTE SPACE ¹	Deep Resolve f-Dixon DANTE SPACE ¹
Deep Resolve Boost & Sharp	No	Yes (research sequence)	Yes (research sequence)
Orientation	Coronal	Coronal	Coronal
Field of view, mm ²	300 × 300	300 × 300	300 × 300
Slice thickness, mm	1	0.9	0.9
No. of slices	120	120	120
Matrix size	352 × 352	352 × 352	352 × 352
Pixel resolution, mm ²	0.85 × 0.85	0.43 × 0.43 (interpolated)	0.43 × 0.43 (interpolated)
TR/TE, ms	2800 / 171	3000 / 251	3000 / 362
TI, ms	250	250	N/A
Inversion RF pulse	Product inversion pulse	C-FOCI	C-FOCI
Bandwidth, Hz/pixel	330	836	862
Turbo factor	130	120	120
Total acceleration factor	4 (CAIPI 2 × 2)	12 (CAIPI 3 × 2, CS 2)	12 (CAIPI 3 × 2, CS 2)
Gadolinium contrast	Enhanced	None	None
Fat suppression	STIR, CHESS	STIR	Fast Dixon
Blood suppression	MSDE	DANTE	DANTE
B ₀ shim	Standard neck	Advanced	Advanced
B ₁ shim	Volume-selective	TrueForm	TrueForm
No. of averages	1	2	2
Total scan time, min:sec	5:22	4:57	4:57

Table 1: Representative 3D MR neurography sequences and acquisition parameters for brachial plexus.



3 Clinical application of deep learning-enhanced SPACE with optimized fat suppression.

Compared to STIR SPACE (3A), Deep Resolve CS CAIPI STIR SPACE with C-FOCI fat suppression (3B) demonstrate improved visualization of the brachial plexus and schwannoma involving the inferior trunk (arrow) with increased detail sharpness and reduced noise, while C-FOCI provides homogeneous fat suppression in the shoulder region. The acquisition time is about 10% shorter than standard STIR SPACE. Both images are MIP reformats with 1 cm thickness.

Sequence	STIR SPACE	Deep Resolve STIR DANTE SPACE ¹	DESS	Deep Resolve DESS ¹	Deep Resolve 3D EPI DESS ¹
Deep Resolve Boost & Sharp	No	Yes (research sequence)	No	Yes (research sequence)	Yes (research sequence)
Orientation	Coronal	Coronal	Coronal	Coronal	Coronal
Field of view, mm ²	400 × 400	400 × 400	400 × 400	400 × 400	400 × 400
Slice thickness, mm	1	1	1	1	1
No. of slices	160	160	160	160	160
Matrix size	384 × 384	384 × 384	448 × 269	400 × 400	400 × 400
Pixel resolution, mm ²	0.52 × 0.52 (interpolated)	0.52 × 0.52 (interpolated)	0.89 × 0.89	0.50 × 0.50	0.50 × 0.50
TR/TE, ms	2800 / 171	2800 / 166	12.27 / 4.41	12.27 / 4.41	13.0 / 4.40
TI, ms	250	250	N/A	N/A	N/A
Flip angle, °	T2 Var	T2 Var	23	23	23
Bandwidth, Hz/pixel	343	352	302	338	962
Turbo factor	130	130	N/A	N/A	5 (EPI factor)
Total acceleration factor	4 (CAIPI 2 × 2)	6 (CAIPI 3 × 2)	None	4 (CAIPI 2 × 2)	4 (CAIPI 2 × 2)
Gadolinium contrast	Enhanced	None	None	None	None
Fat suppression	STIR, CHESS	STIR, CHESS	Water excitation	Water excitation	Fast water excitation
Blood suppression	MSDE	DANTE	N/A	N/A	N/A
B ₀ shim	Standard	Standard	Standard	Standard	Standard
B ₁ shim	TrueForm	TrueForm	TrueForm	TrueForm	TrueForm
No. of averages	1.4	2	1	1	1
Total scan time, min:sec	5:30	4:56	5:13	2:46	2:20

Table 2: Representative 3D MR neurography sequences and acquisition parameters for lumbosacral plexus.

CHESS relies on the resonance frequency difference between fat and water, but it becomes unreliable in regions with B_0 inhomogeneity, such as the brachial plexus and lumbosacral plexus, or in off-center extremity imaging. STIR remains one of the most robust fat suppression methods because it nulls fat based on the T1 relaxation time rather than spectral separation, making it insensitive to B_0 inhomogeneity. However, conventional STIR has disadvantages including lower signal-to-noise ratio and longer scan times.

In optimized 3D STIR SPACE, a C-shaped frequency offset corrected inversion (C-FOCI) pulse improves fat suppression by providing relatively uniform fat inversion across both B_0 and B_1 inhomogeneity [9]. This is particularly relevant in the neck–shoulder region, where field inhomogeneity commonly impairs uniform fat suppression. Combined with DL reconstruction, C-FOCI STIR SPACE achieves high-resolution imaging with robust fat suppression and clinically acceptable scan times (Fig. 3).

Dixon imaging is another widely used fat suppression strategy in MR neurography. It separates water and fat based on their precession frequency differences and provides homogeneous fat suppression with minimal

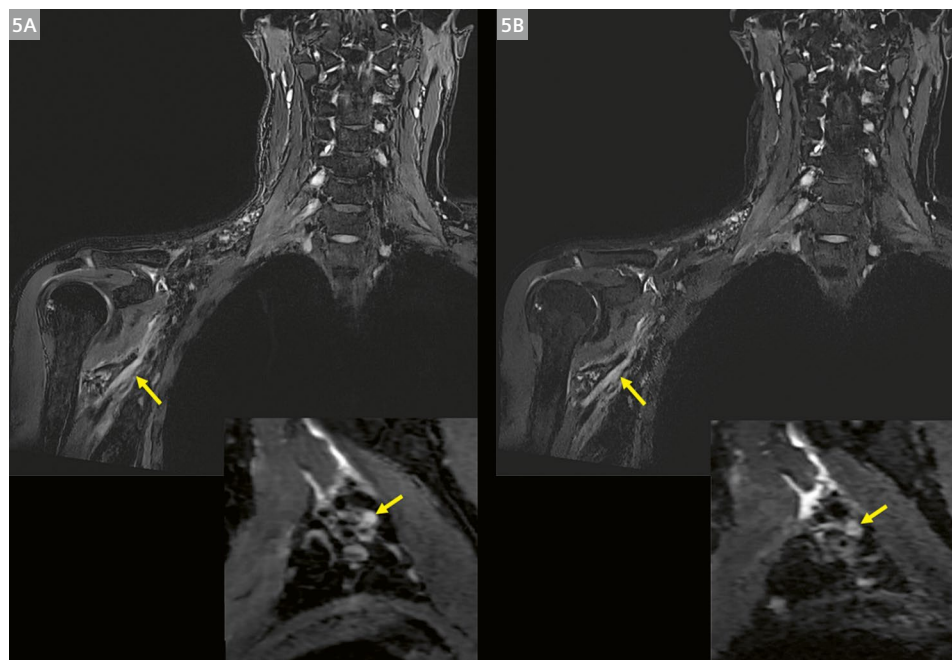
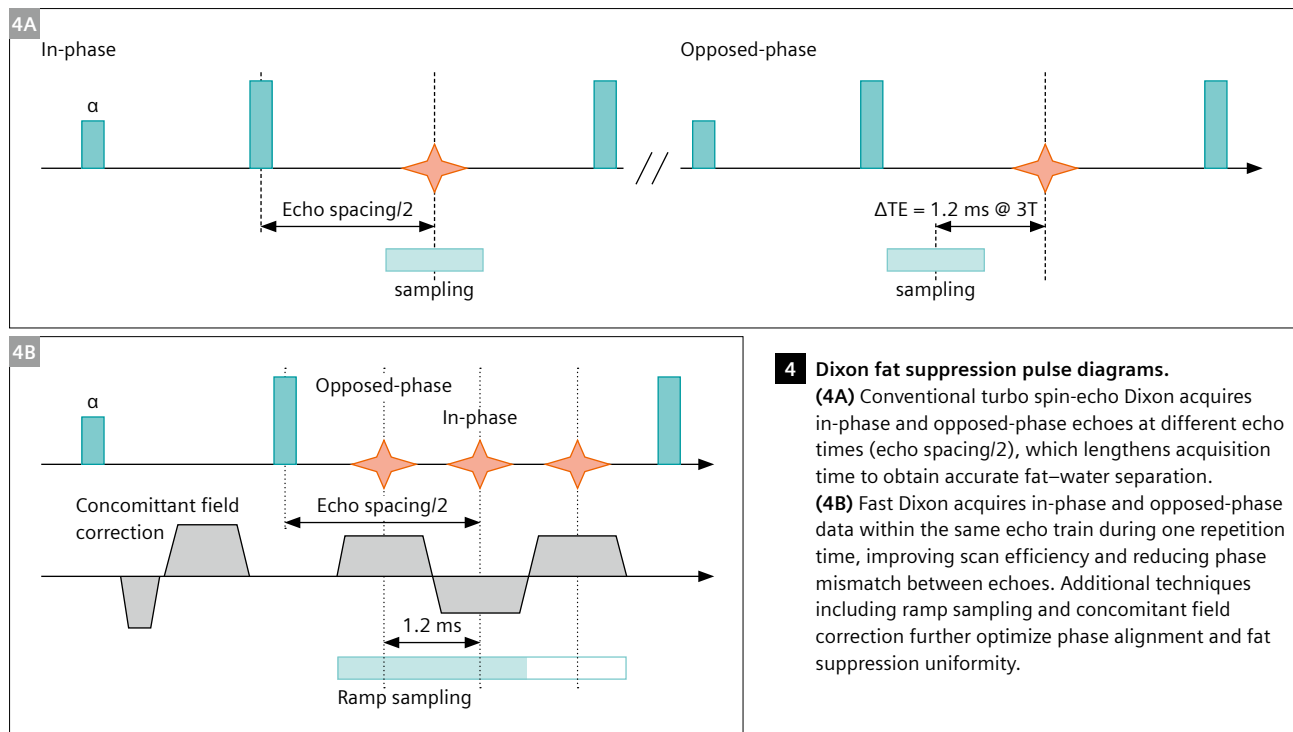
signal-to-noise penalty. Because Dixon is relatively insensitive to B_0 and B_1 inhomogeneity, it is particularly useful for large field-of-view plexus imaging and off-center extremity imaging [10].

Conventional turbo spin-echo Dixon imaging requires echoes that are acquired at different echo times or in separate acquisitions, which may increase scan time and introduce phase errors between echoes during motion. These phase mismatches can result in incomplete fat suppression or water–fat swapping, particularly in regions affected by respiratory or vascular motion. Fast Dixon (f-Dixon) overcomes these limitations by acquiring opposed-phase and in-phase data within the same echo train [11]. This approach improves scan efficiency and reduces phase mismatch between echoes (Fig. 4). Additional strategies, such as ramp sampling and concomitant field correction, further improve acquisition efficiency and phase alignment.

In recent SPACE development, f-Dixon provides robust fat suppression directly integrated into the 3D acquisition. f-Dixon SPACE offers more homogeneous fat suppression compared to STIR while preserving brighter nerve signal without prolonging scan time (Fig. 5).

Sequence	DESS	Deep Resolve DESS ¹	Deep Resolve 3D EPI DESS ¹
Deep Resolve Boost & Sharp	No	Yes (research sequence)	Yes (research sequence)
Orientation	Coronal	Coronal	Coronal
Field of view, mm ²	200 × 200	200 × 200	200 × 200
Slice thickness, mm	1	1	1
No. of slices	112	112	112
Matrix size	192 × 192	192 × 192	192 × 192
Pixel resolution, mm ²	1.04 × 1.04	0.52 × 0.52 (interpolated)	0.52 × 0.52 (interpolated)
TR/TE, ms	12.27 / 4.41	12.27 / 4.41	18.0 / 4.80
Flip angle, °	23	23	23
Bandwidth, Hz/pixel	338	338	500
EPI factor	N/A	N/A	10
Total acceleration factor	None	4 (CAIPI 2 × 2)	4 (CAIPI 2 × 2)
Gadolinium contrast	None	None	None
Fat suppression	Water excitation	Water excitation	Fast water excitation
B_0 shim	Standard	Standard	Standard
B_1 shim	TrueForm	TrueForm	TrueForm
No. of averages	1	1	1
Total scan time, min:sec	5:29	1:45	1:12

Table 3: Representative 3D MR neurography sequences and acquisition parameters for extremity regions (knee/elbow).



- 5** **Comparison of fat suppression techniques in radial neuropathy (arrow).** Deep Resolve f-Dixon SPACE¹ (**5A**) and Deep Resolve STIR SPACE¹ (**5B**) demonstrate comparable spatial resolution and fat suppression quality. f-Dixon preserves superior nerve signal intensity compared to STIR without increasing acquisition time, providing improved visualization of affected nerve segments, particularly evident on oblique sagittal reformatted images.

(3) Vascular suppression requirements

Vascular suppression is critical in MR neurography, as slow-flowing venous structures produce high-signal linear appearances that can obscure or mimic peripheral nerves — which is particularly challenging in the pelvis and extremities, where nerves course adjacent to abundant vascular structures [1]. In the brachial plexus, small branch nerves are closely associated with subclavian, axillary, and cervical vessels [12].

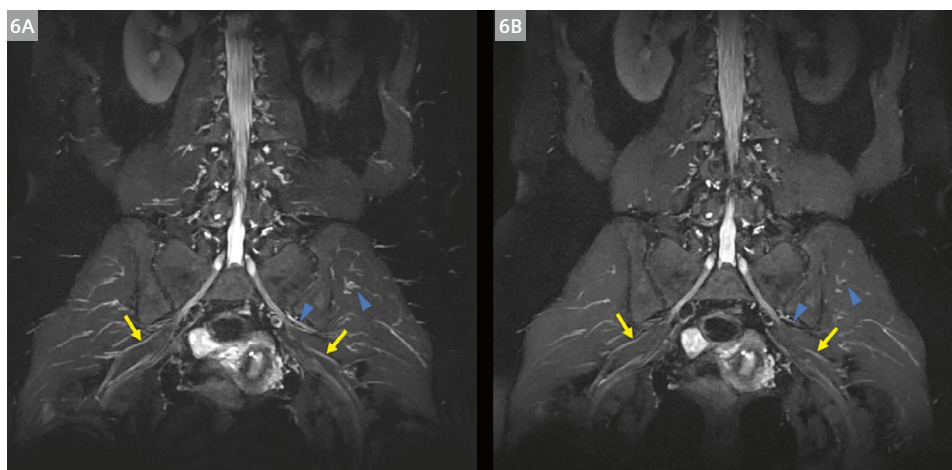
Contrast-enhanced 3D STIR SPACE remains a robust vascular suppression technique. After gadolinium administration, the shortened T1 relaxation time of blood approaches that of fat, allowing STIR to suppress both fat and vascular signal simultaneously. This improves visualization of small branch nerves and the maximum intensity projection (MIP) image quality [12]. However, this approach has limitations: It obscures rather than highlights lesions with blood–nerve barrier disruption, and requires gadolinium administration.

Non-contrast black-blood preparation techniques offer an alternative option: Motion-sensitized driven equilibrium (MSDE) uses motion-sensitizing gradients to dephase flowing spins, but its performance may be limited by signal-to-noise ratio loss, B_1 sensitivity, motion artifacts, and image blurring [13]. As an alternative, delay alternating with nutation for tailored excitation (DANTE) employs trains of low flip-angle pulses to attenuate flowing spins with more effective suppression of slow flow [14]. In lumbosacral plexus imaging, DANTE-prepared STIR SPACE demonstrates superior suppression of slow-flowing venous structures with improved visualization of nerve anatomy [2] (Fig. 6).

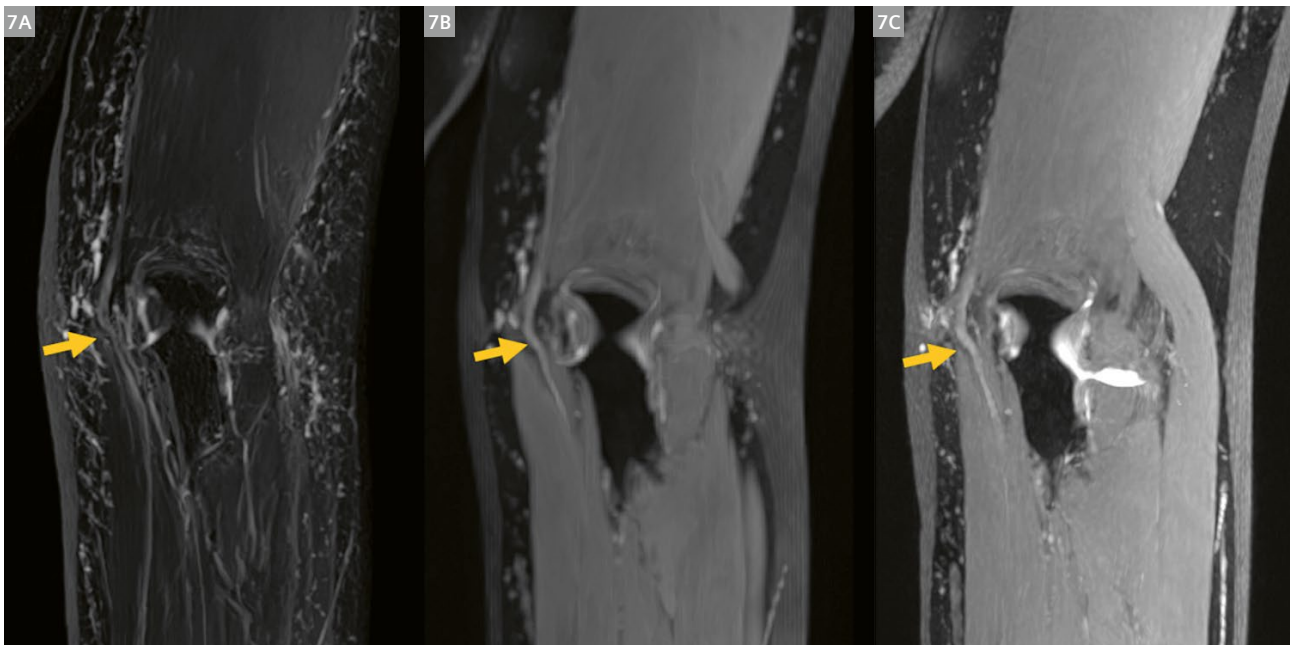
DESS: An alternative to 3D SPACE sequences for MR neurography

Diffusion-weighted reversed fast imaging with steady-state precession (DW-PSIF) has been a favored gradient-echo-based MR neurography technique because peripheral nerves appeared relatively bright due to its heavily T2-weighted contrast and diffusion gradient-based vascular suppression [15]. However, dual echo steady state (DESS) is increasingly used as an alternative, providing high-resolution 3D coverage with improved image quality and intrinsic vascular suppression.

DESS is a gradient-echo steady-state sequence that generates images from the combination of two echoes: a free induction decay component and a spin-echo-like component. Compared with STIR SPACE, DESS generally provides lower nerve-to-background contrast; normal nerves may be less conspicuous because their signal can be similar to adjacent soft tissue. Therefore, DESS is not ideal when the primary goal is to make normal nerves appear bright against a uniformly suppressed background. However, DESS offers two important advantages for MR neurography: efficient fat suppression using water excitation, and intrinsic vascular suppression related to its gradient-echo steady-state mechanism. Moving blood spins experience phase mismatch between the two echoes, whereas relatively stationary nerve tissue maintains phase coherence. Additionally, the short T2* relaxation of deoxy-hemoglobin in venous blood contributes to venous signal loss [2]. These mechanisms make DESS particularly relevant for lumbosacral plexus and extremity peripheral nerve imaging, where STIR SPACE still fails to completely suppress vessel signals (Fig. 7).

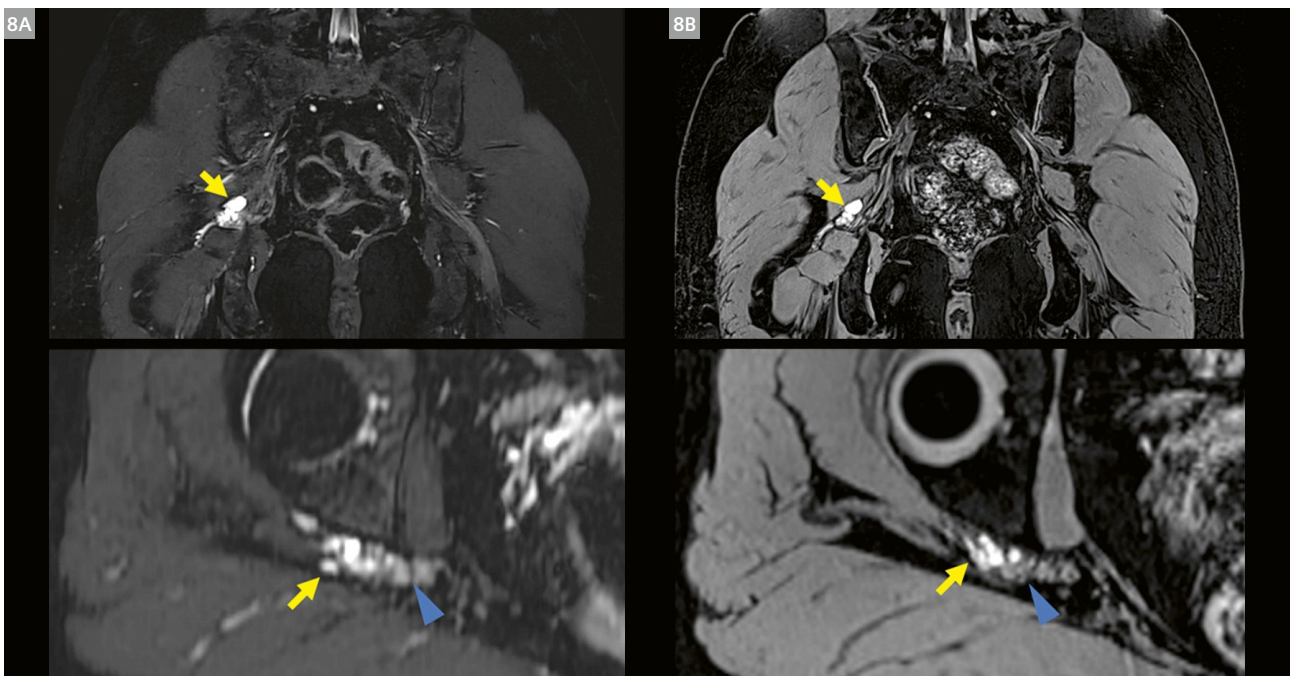


6 Comparison of vascular suppression techniques in non-contrast lumbosacral plexus (LSP) MRN. (6A) MSDE-prepared STIR SPACE¹ and (6B) DANTE-prepared STIR SPACE¹ demonstrate different vascular suppression performance. DANTE provides superior suppression of slow-flowing venous structures, particularly the superior gluteal vein and muscular veins (arrowheads), compared to MSDE. Both images successfully depict sciatic nerve anatomic variants — the left sciatic nerve demonstrates Beaton and Anson type B variant, while the right sciatic nerve shows the type C variant (arrows).



7 Comparison of vascular suppression and image quality in ulnar neuropathy.

Coronal MIP reformats (1 cm thickness) of three different sequences: **(7A)** f-Dixon DANTE-prepared SPACE (4:33 min), **(7B)** DESS (5:18 min), and **(7C)** Deep Learning-enhanced DESS (1:31 min). Both DESS sequences (7B, 7C) demonstrate superior vascular suppression compared to SPACE (7A), with effectively suppressed vessel signals and clear visualization of the affected neuropathy segment. DL-enhanced DESS (7C) achieves significantly shorter acquisition time while providing markedly improved signal-to-noise ratio and image sharpness compared to conventional DESS.



8 Deep Resolve 3D EPI DESS for lumbosacral plexus imaging.

Compared to STIR SPACE **(8A)**, Deep Resolve 3D EPI DESS¹ **(8B)** provides superior visualization of the relationship between the paralabral cyst (arrow) and right sciatic nerve bundle (arrowheads), with markedly improved conspicuity of individual nerve fascicles. The enhanced sharpness and reduced noise of DL reconstruction enable confident assessment of nerve bundle compression and fascicular involvement, demonstrating the potential clinical value of Deep Resolve DESS.¹

Deep Resolve 3D DESS

DL reconstruction combined with dedicated undersampling strategies such as CAIPIRINHA or CS substantially improves acceleration in DESS [16]. Deep Resolve DESS reduces noise, improves perceived sharpness, and increases confidence in the visualization of small nerve branches [17]. Importantly, acquisition time can be reduced to approximately 25%–50% of conventional DESS protocols (Tables 2 and 3). Although DL reconstruction does not fundamentally change the contrast mechanism of DESS, it effectively enhances nerve visualization through noise reduction and improved image sharpness, allowing better differentiation of normal nerves from adjacent soft tissue. Deep Resolve DESS is especially useful in the lumbosacral plexus or upper extremities, where the magnetic field is inhomogeneous and the complex anatomy makes nerve branches difficult to separate from surrounding muscles and adjoining vessels.

Deep Resolve 3D EPI DESS

CAIPIRINHA-accelerated 3D echo-planar imaging (EPI) substantially reduces acquisition time compared with conventional 3D gradient-echo imaging, with further improvement in image quality through deep learning reconstruction [18]. With appropriate selection of segmentation, a moderate EPI factor, CAIPIRINHA sampling, and multishot design, potential artifacts such as distortion, blurring, and residual phase error can be minimized while preserving the advantages of EPI-based acquisition [19].

By incorporating an echo-planar readout into the DESS framework, EPI DESS maintains DESS-type contrast while potentially enabling shorter scan times, broader anatomic coverage, or higher spatial resolution. Although Deep Resolve 3D EPI DESS has not yet been systematically evaluated in MR neurography, a segmented 3D readout design could potentially benefit lumbosacral plexus and extremity peripheral nerve imaging through improved acquisition efficiency (Fig. 8).

Conclusion

Advanced 3D MR neurography has been transformed through integrated technical innovations that address high spatial resolution, homogeneous fat suppression, and effective vascular suppression. SPACE with DL CS CAIPI acceleration and optimized fat/vascular suppression techniques, combined with complementary DESS-based approaches and deep learning reconstruction, provide efficient, high-quality imaging of plexus and peripheral nerve anatomy.

The integration of these technological advances provides flexible and efficient imaging protocols that substantially improve visualization of peripheral nerve anatomy and pathology. These innovations enhance diagnostic confidence in the evaluation of plexus and peripheral nerve disorders, ultimately improving patient management and clinical outcomes.

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Combined Compressed Sensing-CAIPIRINHA DESS with Deep Learning Reconstruction for Highly Accelerated 7T MR Neurography

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Introduction

Magnetic resonance neurography of the cervical spine greatly benefits from the signal-to-noise ratio available at 7T. The 3D double-echo steady-state (DESS) sequence with water excitation provides excellent visualization of cervical nerve anatomy and intervertebral discs while effectively suppressing surrounding bone and fat tissue [1]. This enables high-contrast images that clearly depict micro-structural detail such as intradural rootlets/roots, dorsal root ganglia (DRGs), and the proximal spinal nerves [2, 3]. A key challenge, however, is the relatively long acquisition time required to achieve the necessary sub-millimeter isotropic resolution. Extended scan durations make the examination more vulnerable to motion artifacts caused by swallowing, breathing, or cerebrospinal fluid pulsation.

Substantial reductions in scan time can be achieved through coherent (e.g., GRAPPA, CAIPIRINHA) [4, 5] and incoherent (compressed sensing [CS]) undersampling [6] combined with deep learning (DL) reconstruction. Combining CAIPIRINHA and CS with DL reconstruction may enable even greater acceleration while preserving or improving image quality [7] compared to state-of-the-art CAIPIRINHA or CS.

We explored a novel joint CS + CAIPIRINHA (CS-CAIPI) data sampling scheme with DL reconstruction for highly accelerated 3D DESS MR neurography¹ at 7T.

Parameter	DESS
TR/TE (ms)	8.0 / 3.0
Flip angle (°)	15
Receiver bandwidth (Hz/Px)	434
Number of slices	384
Reconstructed voxel size (mm ³)	0.35 × 0.35 × 0.35
FOV (mm ²)	180 × 180
Acquisition matrix	256 × 256
Fat-water contrast	water excitation
Phase oversampling (%)	40
Phase encoding direction	right-left
Imaging plane orientation	coronal

Table 1: Acquisition parameters for the 7T double-echo steady-state sequence (DESS).

Acquisition parameters for the 3D DESS sequence were kept constant across acceleration factors and undersampling schemes.

FOV = field of view; TE = echo time; TR = repetition time

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.

Material and methods

Image acquisition and reconstruction

Two healthy volunteers underwent 7T cervical spine MRI on the MAGNETOM Terra.X (Siemens Healthineers, Erlangen, Germany) using an 8Tx/8Rx neck coil (Rapid Biomedical, Rimpfing, Germany).

A 3D DESS sequence with conventional 2-fold GRAPPA acceleration served as the reference. The sequence was then repeated with a research application sequence¹ featuring CAIPIRINHA-only undersampling (DL-CAIPIRINHA) and combined CAIPIRINHA + CS undersampling (DL-CS-CAIPI) with DL reconstruction, while keeping all other parameters identical (Table 1). Acceleration factors were progressively increased (DL-CAIPIRINHA: R = 2–12; DL-CS-CAIPI: R = 4–16).

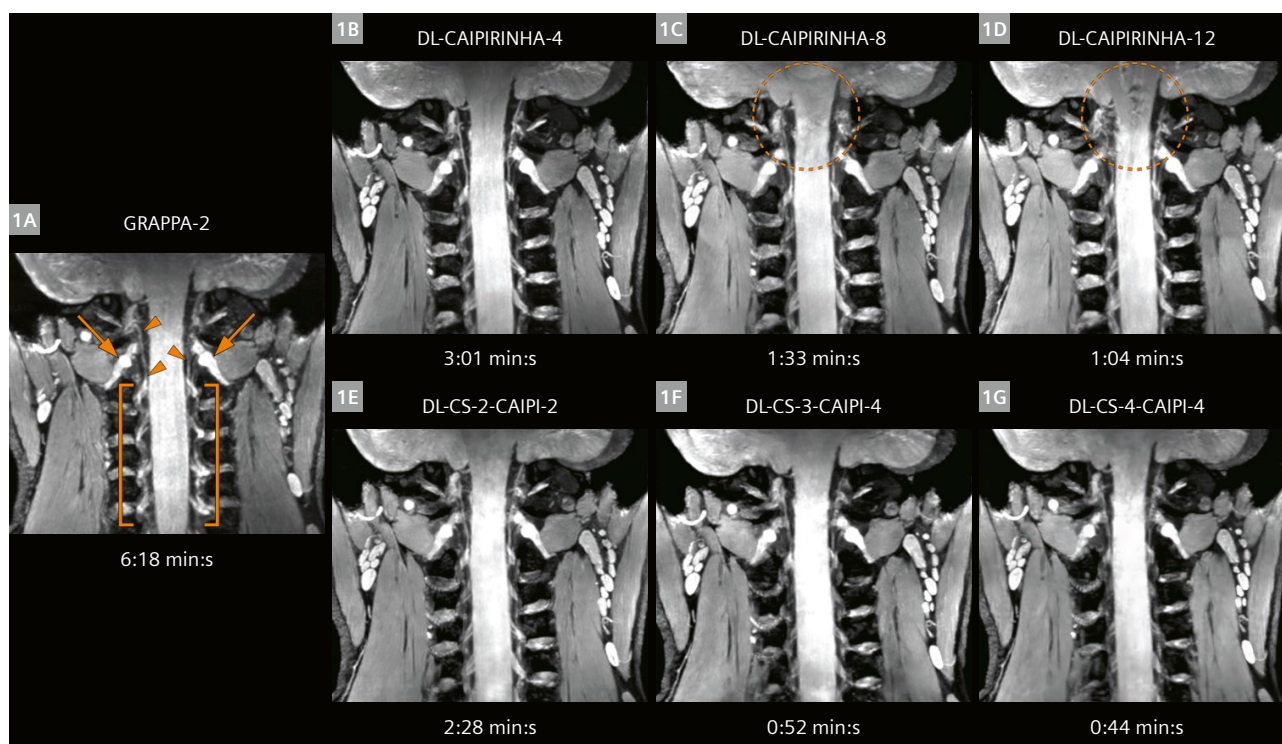
The combined undersampling scheme first applies a regular CAIPIRINHA pattern, then superimposes a variable-density Poisson-disc (CS) mask. Total acceleration is the product of the two factors ($R_{\text{Total}} = R_{\text{CAIPIRINHA}} \times R_{\text{CS}}$). The reconstruction uses a two-stage pipeline: initial CS

reconstruction with redundant Haar wavelets (FISTA optimization) [8] followed by DL-based refinement [9] inspired by variational networks (trained on fully sampled 1.5T/3T data and fine-tuned on 7T data). Acquisition times decreased from 6:18 minutes (GRAPPA-2) to only 44 seconds with 16-fold DL-CS-CAIPI.

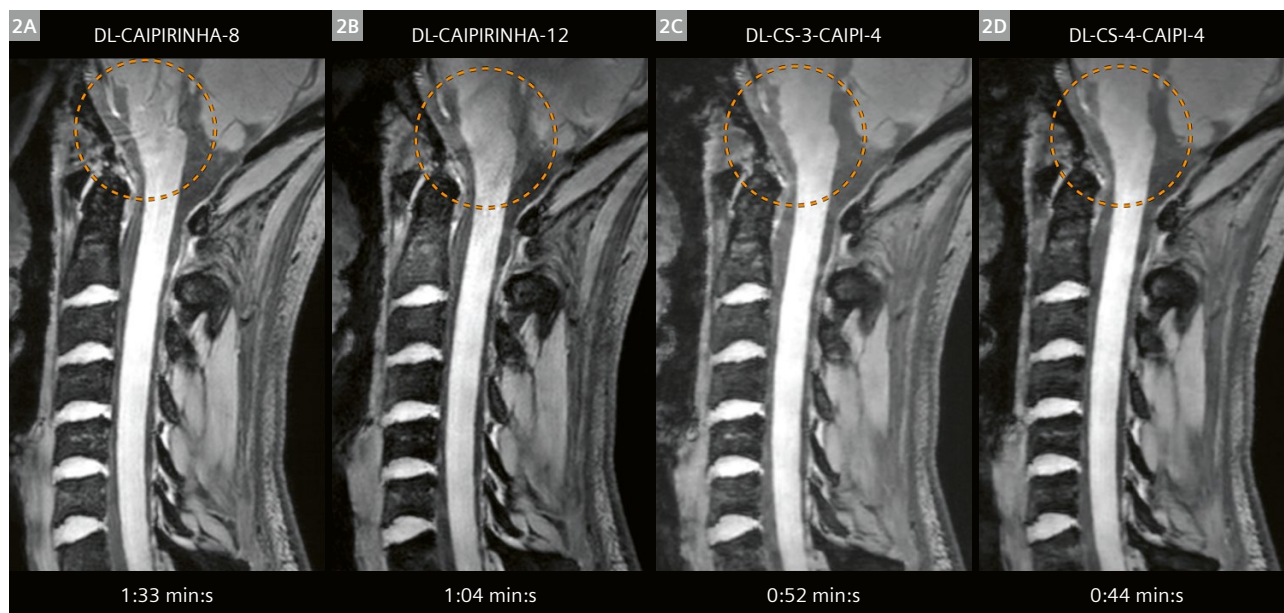
Images were reviewed with a focus on overall quality, noise, artifacts, and sharpness of neural structures.

Results

Figures 1 and 2 show representative images from two healthy volunteers, illustrating the performance of the different undersampling schemes across a range of acceleration factors. Optimal image quality and neural-structure sharpness were observed with 4-fold DL-CAIPIRINHA compared with the standard GRAPPA reference, with a 50% shorter scan time and markedly reduced motion artifacts.



1 Coronal 3 mm maximum intensity projections of 3D DESS images acquired using 2-fold GRAPPA (1A), DL-CAIPIRINHA (1B–D), and combined DL-CS-CAIPI (1E–G) at progressively higher acceleration factors with their corresponding acquisition times. The 2-fold GRAPPA image (1A) depicts the bilateral C2 dorsal root ganglia (arrows), C3–C6 intraspinal rootlets/roots (brackets), and spinal roots of the spinal accessory nerve (arrowheads). Neural-structure sharpness is improved with 4-fold DL-CAIPIRINHA (1B). At higher CAIPIRINHA-only accelerations (R = 8 and R = 12 in 1C and 1D), streak-like aliasing artifacts become visible (circles). The combined DL-CS-CAIPI technique maintains good overall anatomic detail, although some blurring appears at R = 12 and R = 16 (1F, 1G).



2 Sagittal 3D DESS images acquired using DL-CAIPIRINHA at $R = 8$ and $R = 12$ (**2A**, **2B**) and combined DL-CS-CAIPI at $R = 12$ and $R = 16$ (**2C**, **2D**) with their corresponding acquisition times. Streak-like aliasing artifacts around the brainstem (circles in **2A** and **2B**) compromise image quality with DL-CAIPIRINHA alone. In contrast, the combined DL-CS-CAIPI approach shows only a mild and less disruptive “leopard-like” speckled pattern at the brainstem (circles in **2C** and **2D**), even at the extreme acceleration of $R = 16$.

At higher accelerations (particularly at $R \geq 8$), however, standalone DL-CAIPIRINHA suffered from increasing streak-like aliasing artifacts, pronounced around the brainstem and image periphery.

The combined DL-CS-CAIPI DESS proved more robust at acceleration factors beyond $R = 8$. Image quality remained comparable to GRAPPA-2 until 12-fold DL-CS-3-CAIPI-4, despite a 6-fold reduction in scan time. Although combined DL-CS-CAIPI schemes demonstrated higher baseline reconstruction artifact levels (mainly blurring, speckled pattern), they had fewer severe reconstruction artifacts at $R = 12$ and $R = 16$ compared with DL-CAIPIRINHA-12. The main trade-off for DL-CS-CAIPI at high acceleration was increased blurring beyond $R = 8$, leading to reduced edge sharpness of neural structures compared with standalone DL-CAIPIRINHA at equivalent total acceleration.

Conclusion

Coherent CAIPIRINHA undersampling combined with DL reconstruction delivered superior image quality in 7T MR neurography while halving acquisition time compared with conventional GRAPPA. The addition of incoherent CS undersampling enabled high accelerations ($R \geq 8$) with a more favorable artifact profile, replacing the structured aliasing artifacts seen in CAIPIRINHA-only acceleration with subtler, less disruptive “leopard-like” patterns [10].

The novel integration of coherent and incoherent undersampling in a combined CS-CAIPI framework with DL reconstruction provides a robust approach for preserving image quality at acceleration factors beyond $R = 8$. It enables efficient high-resolution cervical spine MRI with up to a 6-fold reduction in scan time.

In summary, our initial experience suggests that the synergistic combination of CS-CAIPI and DL reconstruction makes highly accelerated, high-resolution 7T MR neurography of the cervical spine feasible while preserving anatomic detail.

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Radial UTE MRI in Musculoskeletal Diseases: Current Clinical Applications and Future Perspectives

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Introduction

Magnetic resonance imaging (MRI) is the modality of choice for evaluating most soft-tissue structures in the musculoskeletal system because of its excellent soft-tissue contrast and lack of ionizing radiation. However, conventional MRI sequences typically use echo times longer than one millisecond and therefore fail to capture signal from tissues with ultrashort transverse relaxation times, such as cortical bone, calcified cartilage, fibrocartilage, the deep or enthesial portions of tendons and ligaments, and cartilaginous endplate at the discovertebral junction [1, 2]. Because these short-T2 tissues are frequently involved in musculoskeletal disease, conventional MRI leaves a diagnostic gap that has traditionally been addressed by radiography or computed tomography (CT).

Three-dimensional (3D) radial ultrashort echo time (UTE)* MRI helps bridge this gap by sampling the free induction decay within tens of microseconds after non-selective excitation, using a center-out radial *k*-space trajectory [3]. This technique enables visualization of tissues that are normally signal-void on conventional MRI, and can provide CT-like contrast for cortical bone and mineralized tissue while preserving the soft-tissue advantages of MRI. As a result, radial UTE MRI has emerged as a promising adjunct to standard musculoskeletal MRI protocols, particularly when simultaneous assessment of bone, mineralized tissue, marrow, and soft-tissue structures is clinically desirable.

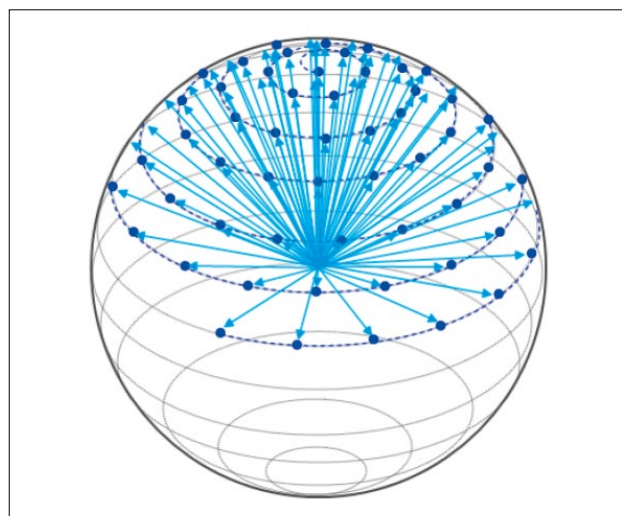
Technical principles of radial UTE MRI

Three-dimensional ultrashort echo time (3D UTE) imaging was performed using a center-out radial research application sequence with TE 40 μ s [3]. A short nonselective radiofrequency (RF) block pulse (40 μ s) was followed by immediate data sampling after transmit/receive switching. To minimize TE, data acquisition was initiated during the

ramp-up of the readout gradient. *k*-space was sampled from the center to the periphery using an isotropic 3D radial “kooshball” trajectory (Fig. 1). Image reconstruction was performed via a non-uniform fast Fourier transform using Cartesian gridding with a Kaiser-Bessel window function. To reduce trajectory deviations, the *k*-space locations of the samples were estimated using a gradient impulse response function [4]. The imaging parameters are summarized in Table 1.

Clinical applications

The clinical value of 3D radial UTE MRI lies in its ability to depict cortical bone and mineralized tissue with CT-like contrast while retaining the soft-tissue information available from conventional MRI. In current musculoskeletal imaging practice, UTE is best understood as an additional sequence within a standard MRI protocol, rather than a stand-alone examination. The following sections summarize representative clinical applications in which UTE MRI may provide incremental diagnostic value.



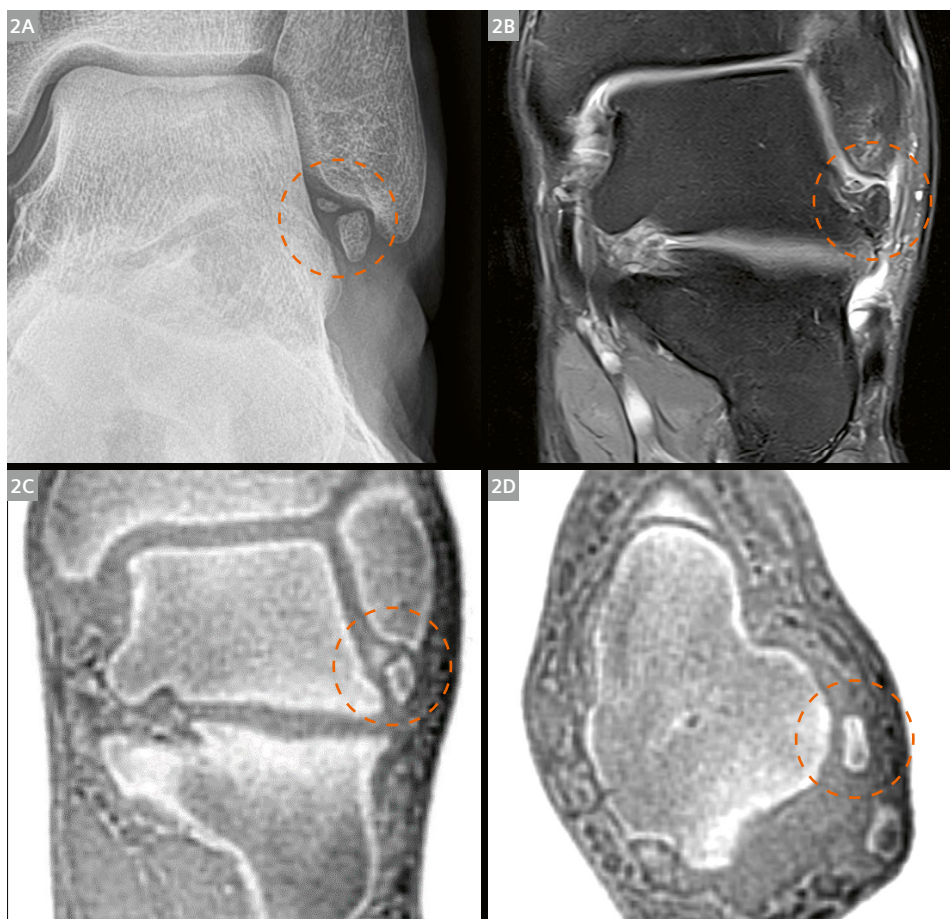
1 Schematic visualization of the radial sampling scheme.

*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.

Bone morphology and cortical contour assessment

Accurate depiction of cortical contour and osseous morphology is fundamental in many musculoskeletal diagnoses [5]. On conventional MRI, cortical bone is usually seen as signal-void structure, which can make subtle cortical irregularity, osseous remodeling, or complex bony anatomy difficult to assess. Radial UTE MRI can improve visualization of cortical margins and bone morphology by providing CT-like contrast within an MRI examination [6, 7].

This capability is particularly useful in joints and anatomic regions where both osseous morphology and soft-tissue status are clinically relevant. For example, in the spine and appendicular skeleton, UTE MRI can depict cortical contour abnormalities while conventional MRI sequences demonstrate marrow edema, disc disease, ligamentous injury, synovitis, or soft-tissue pathology (Fig. 2). Thus, UTE MRI may complement routine MRI by integrating osseous and soft-tissue information in a single radiation-free examination (Fig. 3) [7, 8].



2 Symptomatic bipartite os subfibulare in a 22-year-old male with chronic ankle instability. Anteroposterior ankle radiograph (**2A**), coronal fat-suppressed T2-weighted MR image (**2B**), and coronal and axial UTE MR images (**2C**, **2D**) demonstrate a bipartite os subfibulare. The ossicles (dotted circles) are more clearly delineated with CT-like cortical contrast on UTE than on the corresponding conventional MR image. The patient subsequently underwent surgical treatment.



3 Femoroacetabular impingement (cam type) in a 51-year-old male. Dunn-view radiograph (**3A**) and axial UTE MR image (**3B**) of the left hip demonstrate an osseous prominence (arrows) at the femoral head-neck junction, which is clearly delineated with CT-like cortical contrast on UTE.

Fractures

Fracture evaluation is another important application of radial UTE MRI. Fracture lines, cortical disruption, and displaced osseous fragments are often defined by cortical bone, which is poorly visualized on conventional MRI but can be depicted more directly with UTE-based imaging [7, 8]. UTE MRI may therefore improve confidence in identifying cortical breaks or small osseous fragments when conventional MRI findings are equivocal.

This approach may be especially advantageous when radiation avoidance is desirable, such as in children, young adults, pregnant patients when appropriate, or patients requiring repeated follow-up examinations (Fig. 4) [6, 9]. In addition, UTE MRI may be useful when both fracture morphology and associated marrow or soft-tissue injury need to be assessed. However, because CT still provides superior spatial resolution, UTE MRI should currently be regarded as complementary to CT rather than a universal substitute.

Focal bone lesions

In focal bone lesions, radial UTE MRI can add information about cortical integrity and matrix mineralization to the marrow and soft-tissue information provided by conventional MRI [9, 10]. Conventional MRI is highly sensitive for marrow replacement, edema, soft-tissue extension, endosteal scalloping, and aggressive features. However, it is less specific for mineralized matrix, cortical detail, and small calcified or ossified components.

UTE MRI may help depict patterns of mineralization, including chondroid calcification, osteoid matrix, or nidus mineralization, while also demonstrating cortical

involvement and associated pathologic fracture (Fig. 5). This information can be useful in lesion characterization, treatment planning, and follow-up. Nevertheless, CT remains superior when very fine mineralized matrix detail or subtle cortical destruction must be assessed with maximum spatial resolution.

Joint pathology

Radial UTE MRI may also contribute to the evaluation of degenerative, inflammatory, and infectious joint diseases by improving visualization of cortical and subchondral osseous abnormalities. These include osteophytes, subchondral sclerosis, erosions, osteolysis, and frank bony destruction [7, 8].

Parameters	3D radial UTE sequence
Radial view	40000
TR / TE (ms)	3.9 / 0.04
FOV (mm)	More than 200 mm
Flip angle (deg)	1.5°
Matrix	202 × 202
Slice thickness (mm)	0.7
Slices	288 slices per slab
Bandwidth (Hz/pixel)	1157
Acquisition time (min:sec)	5:13

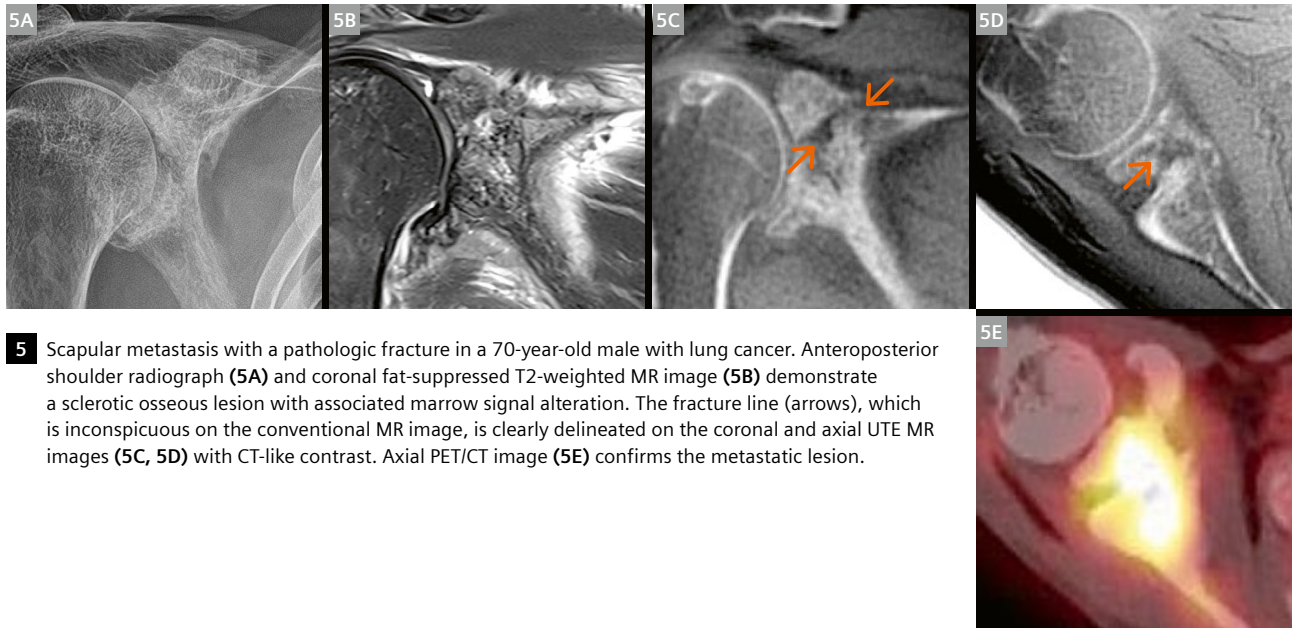
Table 1: 3T MAGNETOM Vida imaging parameters



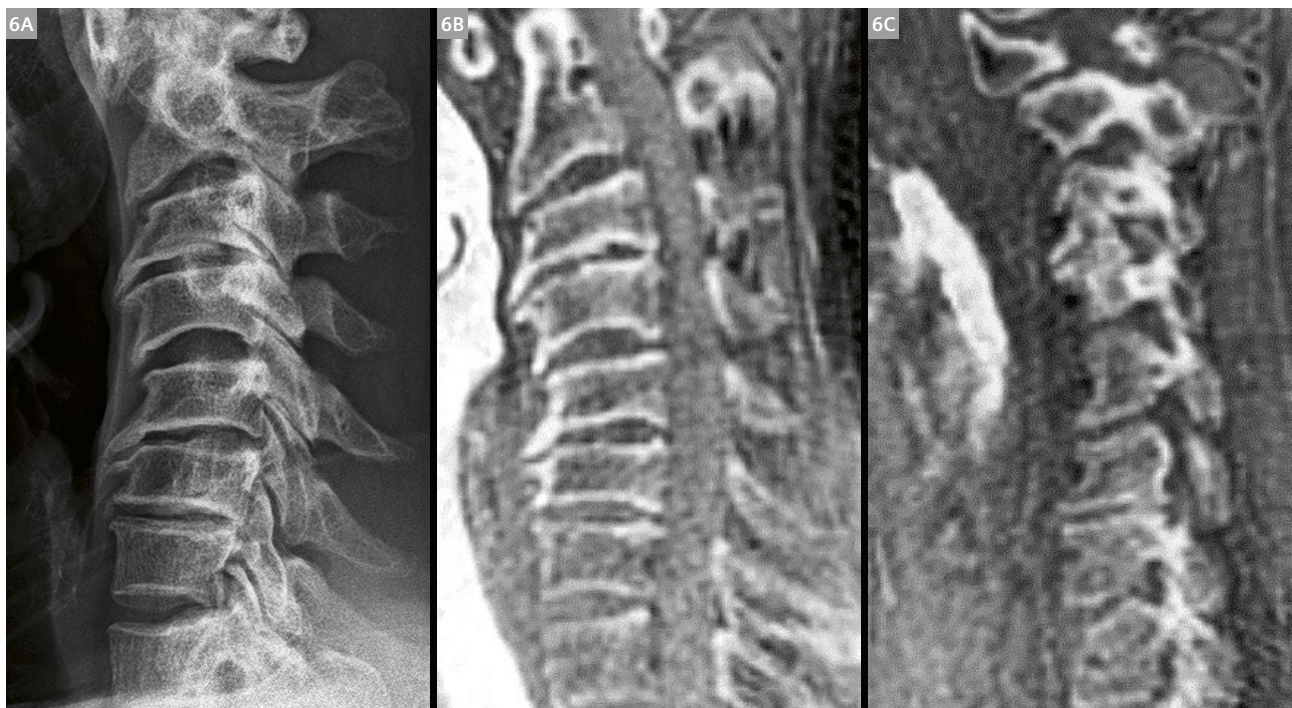
4 Tibial tuberosity avulsion fracture in a 13-year-old boy. Lateral knee radiograph (4A) and sagittal and coronal UTE MR images (4B, 4C) demonstrate a tibial tuberosity avulsion fracture extending through the proximal tibial physis and epiphysis with intra-articular extension. UTE MR images clearly delineate the fracture line and displaced fragment with CT-like cortical contrast.

In degenerative disease in the joints and spine, productive osseous changes such as osteophytes and subchondral remodeling are important markers of disease severity (Fig. 6). In this setting, UTE MRI may help provide a more comprehensive evaluation of both soft-tissue and osseous components of degenerative disease [7, 8].

In inflammatory arthritis, cortical erosions are clinically important because they reflect structural joint damage. UTE MRI may improve depiction of erosive change by better delineating cortical margins. Similarly, in infectious arthritis, early recognition and accurate mapping of cortical erosion or subchondral destruction are important



5 Scapular metastasis with a pathologic fracture in a 70-year-old male with lung cancer. Anteroposterior shoulder radiograph (5A) and coronal fat-suppressed T2-weighted MR image (5B) demonstrate a sclerotic osseous lesion with associated marrow signal alteration. The fracture line (arrows), which is inconspicuous on the conventional MR image, is clearly delineated on the coronal and axial UTE MR images (5C, 5D) with CT-like contrast. Axial PET/CT image (5E) confirms the metastatic lesion.



6 Cervical spondylosis in a 56-year-old male. Lateral cervical spine radiograph (6A) and sagittal UTE MR images (6B, 6C) demonstrate multilevel osteophytes, endplate sclerosis, and disc space narrowing. UTE clearly delineates these osseous degenerative changes with CT-like contrast.

for assessing disease severity and planning treatment. By combining UTE depiction of cortical destruction with conventional MRI findings such as joint effusion, synovitis, abscess, and marrow edema, MRI may provide a more complete radiation-free assessment of joint infection [6–8].

Soft-tissue calcification and ossification

Calcified and ossified soft-tissue deposits have short-T2 properties and may be difficult to characterize on conventional MRI, where they often appear as nonspecific low-signal foci. Radial UTE MRI can directly depict mineralized tissue and may therefore improve detection and characterization of soft-tissue calcification or ossification [8, 9].

This application is relevant in conditions involving calcific tendinopathy, heterotopic ossification, enthesopathic change, and other mineralized deposits within ligaments or periarticular soft tissues (Fig. 7). By clarifying whether a low-signal abnormality on conventional MRI represents mineralization, fibrosis, susceptibility artifact, or another process, UTE MRI may improve diagnostic specificity and help define disease extent [9].

Postoperative evaluation

Postoperative musculoskeletal imaging often requires assessment of both hardware and adjacent bone. Conventional MRI may be limited by susceptibility artifacts near metallic implants, while radiographs provide limited soft-tissue and marrow information. The very short echo time of UTE sequences can reduce susceptibility-related signal loss and distortion, and radial UTE-based approaches may improve visualization of bone and hardware-adjacent structures (Fig. 8) [11].

Advanced applications

Beyond morphologic imaging, UTE-based techniques enable quantitative assessment of short-T2 tissues. Quantitative UTE methods can estimate parameters such as T2*, T1, T1ρ, magnetization transfer ratio, macromolecular proton fraction, and porosity-related indices. These biomarkers are being investigated for evaluating cortical bone quality, tendon and ligament composition, cartilage degeneration, and other tissue-level changes that are difficult to assess with conventional MRI [2, 6, 8, 12–14].

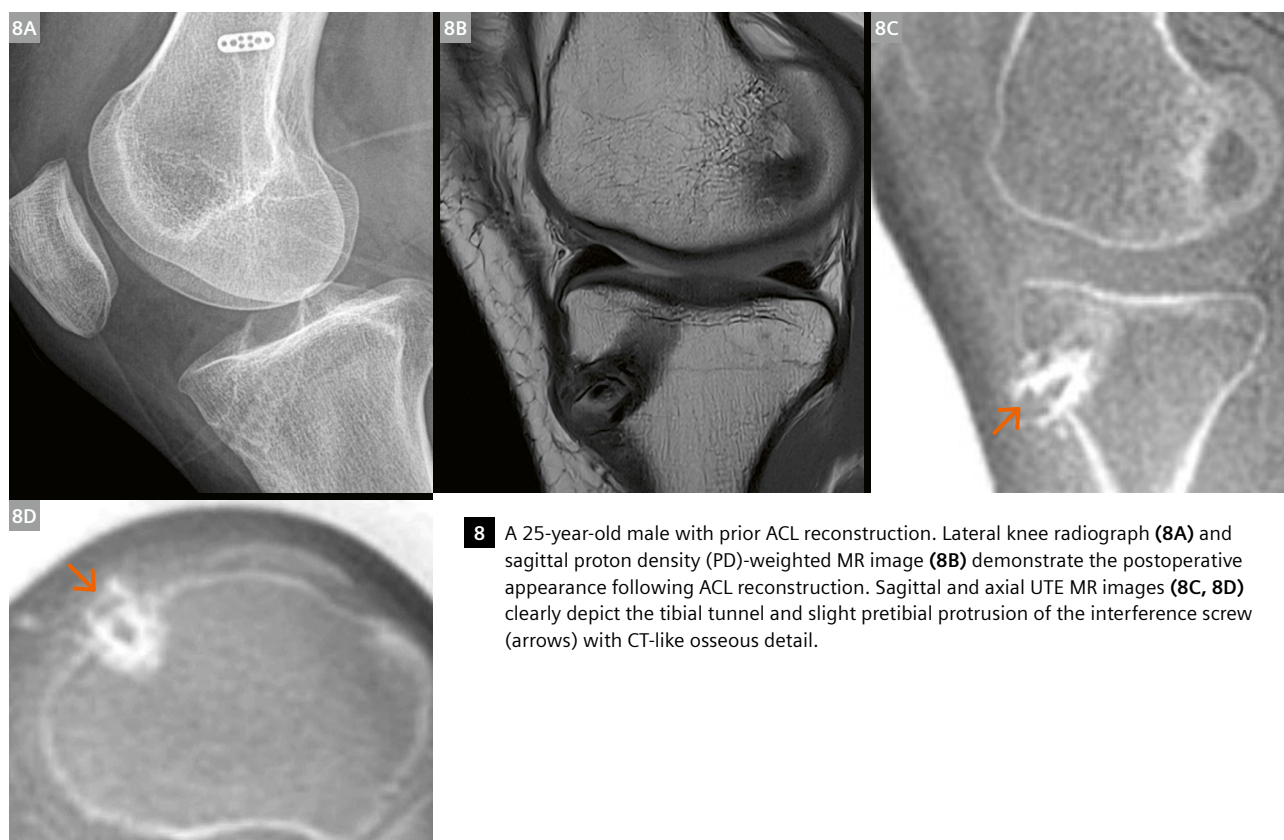
Emerging UTE-based approaches, including quantitative susceptibility mapping, conductivity mapping, and accelerated 3D radial acquisitions with deep learning reconstruction, may further expand the role of UTE MRI in musculoskeletal imaging. These methods remain largely investigational, but they suggest that UTE MRI may eventually contribute not only to structural visualization but also to quantitative tissue characterization.

Limitations and future directions

Despite its advantages, radial UTE MRI has several important limitations. First, although UTE can provide CT-like contrast, it does not yet match the spatial resolution, speed, or robustness of CT for fine cortical detail and subtle mineralization [7, 8, 13]. Second, radial UTE MRI may require longer acquisition times than conventional sequences, increasing susceptibility to patient motion. Nevertheless, the radial nature of the sequence is less susceptible to motion due to repeated sampling of the *k*-space center. Third, radial acquisitions can produce specific artifacts, including streak artifacts, blurring, chemical shift, and artifacts related to gradient imperfections. These artifacts may mimic or obscure pathology



7 Calcific tendinitis of the supraspinatus in a 75-year-old female. The calcific deposit (arrows) is clearly depicted on the 30° caudal tilt radiograph (7A) and the inverted grayscale UTE MR image (7B), which provides CT-like contrast; it appears ambiguous on the corresponding proton density (PD)-weighted MR image (7C).



8 A 25-year-old male with prior ACL reconstruction. Lateral knee radiograph (**8A**) and sagittal proton density (PD)-weighted MR image (**8B**) demonstrate the postoperative appearance following ACL reconstruction. Sagittal and axial UTE MR images (**8C**, **8D**) clearly depict the tibial tunnel and slight pretibial protrusion of the interference screw (arrows) with CT-like osseous detail.

if not recognized. Technical optimization and careful interpretation are therefore essential [12, 15, 16]. Finally, incomplete standardization remains a barrier to broader clinical adoption. However, ongoing advances in accelerated acquisition and image reconstruction, including deep learning-based reconstruction, are improving image quality while reducing scan times. Together with further clinical validation and protocol harmonization, these developments are expected to enhance the clinical feasibility of radial UTE MRI and facilitate its broader integration into routine musculoskeletal imaging across institutions [17].

Conclusion

Radial UTE MRI is a rapidly evolving technique that expands the diagnostic capability of musculoskeletal MRI by enabling visualization of tissues that are poorly seen on conventional sequences. By recovering signal from short-T2 structures and providing CT-like contrast

for cortical bone and mineralized tissue, UTE MRI can complement standard MRI in the evaluation of bone morphology, fractures, focal bone lesions, joint disease, soft-tissue calcification or ossification, and postoperative changes.

At present, radial UTE MRI is best regarded as a powerful adjunct to, rather than a replacement for, CT or conventional MRI. Its clinical role is partially limited by longer acquisition times, sequence-specific artifacts, magic-angle effect, and lower spatial resolution compared with CT. Nevertheless, continuing advances in acquisition and reconstruction are likely to expand the role of radial UTE MRI in comprehensive, radiation-free musculoskeletal imaging.

Acknowledgments

We would like to thank Dr. Sugil Kim of MR Research Collaboration at Siemens Healthineers, Seoul, Korea, for his helpful discussion throughout this study.

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Muhammed Labeeb Kozhikkavil, M.Sc. RT

I grew up in a small village nestled on the outskirts of Kozhikode, a vibrant coastal city on the west coast of Kerala in South India. Growing up, my core interests leaned heavily toward the humanities, history, archeology, and foreign relations. But I also had a deep fascination for physics and how the fundamental laws of nature work.

My path into medical imaging wasn't part of a grand initial plan; it was actually completely accidental. But as Albert Einstein famously said, "I have no special talent. I am only passionately curious." That curiosity kicked in fast for me, turning an unexpected field of study into my lifelong passion. Fresh out of university, I was recruited through campus placement as a graduate trainee engineer with Siemens Ltd. India in the Clinical Application team, covering India and neighboring countries. Later, I moved to China to join the R&D team at Shenzhen SSMR as a protocol developer, before eventually making my way to our global headquarters here in Erlangen.



How did you first come into contact with MRI?

My first real encounter with magnetic resonance imaging was during my studies at the Sree Chitra Tirunal Institute for Medical Sciences and Technology, a renowned tertiary research hospital in India. Seeing how MRI could non-invasively peer inside the human body to help diagnose complex conditions blew my mind. Watching that technology in action sparked a clear realization: I didn't just want to work with these systems; I wanted to be on the front line of developing them.

What do you find motivating about your job?

Today, my work focuses on protocol development and workflow integration for musculoskeletal imaging. I work side by side with pre-development teams and MRI scientists. What motivates me every single day is the direct impact of our work. Knowing that refining an imaging protocol or streamlining a workflow can reduce scan times, improve diagnostic clarity, and make a patient's experience more comfortable keeps me going. It's the perfect blend of high-level science and real-world human empathy.

What are the biggest challenges in your job?

The trickiest and most exciting part of protocol development is translation. MRI scientists and physics teams develop incredible next-generation innovations, while MRI technologists in hospitals need simplicity, speed, and reliability. The biggest challenge is taking those advanced technical developments and seamlessly embedding them

into intuitive, error-free clinical workflows that function effortlessly across diverse global healthcare environments.

What do you think are the most important developments in healthcare?

I believe the shift toward personalized, high-efficiency medicine supported by AI and intelligent automation is the most crucial trend today. In MRI specifically, reducing scan times without compromising image quality is game changing. From my perspective, faster scanning has the potential to improve the MRI experience by helping to shorten examination times and reduce the likelihood of motion-related image-quality limitations/artifacts. The future lies in bridging advanced technology with accessible clinical application.

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

If I had a full month with zero commitments, I would pack a bag and set off on a long cross-border road trip! I have a huge passion for overland travel. There's nothing quite like driving through changing landscapes, crossing borders, and experiencing new cultures at your own pace.

When I'm not dreaming of the open road, I keep active locally. I used to love playing as a goalkeeper on the football pitch until a knee injury sidelined me. These days, you'll find me playing badminton or hitting sixes on the cricket pitch with the local Indian diaspora community here in Erlangen.

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