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Recent Advances in MRI of Articular Cartilage

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Abstract

OBJECTIVE—MRI is the most accurate noninvasive method available to diagnose disorders of articular cartilage. Conventional 2D and 3D approaches show changes in cartilage morphology. Faster 3D imaging methods with isotropic resolution can be reformatted into arbitrary planes for improved detection and visualization of pathology. Unique contrast mechanisms allow us to probe cartilage physiology and detect changes in cartilage macromolecules.

CONCLUSION—MRI has great promise as a noninvasive comprehensive tool for cartilage evaluation.

Keywords

balanced steady-state free precession imaging; bSSFP; cartilage; joint imaging; MRI; osteoarthritis; rapid imaging

Articular cartilage pathology may be the result of degeneration or acute injury. Osteoarthritis is an important cause of disability in our society [1–6] and is marked by degeneration of articular cartilage [7–9]. Acute injury to cartilage can be characterized using MRI [10]. Whether the result is from degeneration or injury, MRI offers a noninvasive means of assessing the degree of damage to cartilage and adjacent bone and of measuring the effectiveness of treatment [11].

Many imaging methods are available to assess articular cartilage. Conventional radiography can be used to detect gross loss of cartilage, evident as narrowing of the distance between the two adjacent bones of a joint [12], but it does not image cartilage directly. Secondary changes such as osteophyte formation can be seen, but conventional radiography is insensitive to early chondral damage. Arthrography, alone or combined with conventional radiography or CT, is mildly invasive and provides information limited to the contour of the cartilage surface [13].

MRI, with its excellent soft-tissue contrast, is the best imaging technique currently available for the assessment of articular cartilage [14–19]. Imaging regions of cartilage damage has the potential to provide morphologic information, such as fissuring and the presence of partial- or full-thickness cartilage defects. Cartilage lesions on MRI are often graded on a modified Outerbridge or Noyes scale, corresponding to arthroscopic grading [20–22]. A common grading scale is shown in Table 1. In addition to morphologic assessment, the many tissue

parameters that can be measured by MRI techniques also have the potential to provide biochemical and physiologic information about the cartilage [23].

An ideal MRI study for cartilage should provide accurate assessment of cartilage thickness and volume, show morphologic changes of the cartilage surface, show internal cartilage signal changes, and allow evaluation of the subchondral bone for signal abnormalities. Also, it would be desirable for MRI to provide an evaluation of the underlying cartilage physiology, including providing information about the status of the glycosaminoglycan (GAG) and collagen matrices. Conventional MRI sequences do not provide a comprehensive assessment of cartilage, lacking either in spatial resolution or specific information about cartilage physiology.

Conventional MRI Methods

MRI has emerged as the leading method of imaging soft-tissue structures around joints [24]. One of the major advantages of MRI is its ability to manipulate contrast to highlight different tissue types. The common contrast mechanisms used in MRI are 2D or multislice T1-weighted, proton density, and T2-weighted imaging with or without fat suppression. Imaging hardware and software have changed considerably over time, including improved gradients and radiofrequency coils, fast or turbo spin-echo imaging, and techniques such as water-only excitation.

Although the tissue relaxation times and imaging parameters are the major determinants of contrast between cartilage and fluid, lipid suppression increases contrast between nonlipid and lipid-containing tissues and affects how the MR scanner sets the overall dynamic range of the image. The most common type of lipid suppression is fat saturation, in which fat spins are excited and then dephased before imaging. Another option is spectral-spatial excitation, in which only water spins in a slice are excited [25]. Finally, in areas of magnetic field inhomogeneity, inversion recovery provides a way to suppress lipids at the expense of the signal-to-noise ratio (SNR) and contrast-to-noise ratio.

The type of contrast used in cartilage imaging is critical to the visibility of lesions and to the SNR of the cartilage itself. Although T2-weighted imaging creates contrast between cartilage and synovial fluid, it does so at the expense of cartilage signal. The high signal from fluid is useful to highlight surface defects such as fibrillation or fissuring, but variation in internal cartilage signal is poorly depicted. Also, these scans are often 2D, leaving a small gap between slices that may miss small areas of cartilage damage.

Three-dimensional spoiled gradient-recalled echo (SPGR) imaging with fat suppression produces high cartilage signal, but low signal from adjacent joint fluid. Currently, this technique is the standard for quantitative morphologic imaging of cartilage [18,22,26]. This technique is useful for cartilage volume and thickness measurements, but does not adequately highlight surface defects with fluid and does not allow thorough evaluation of other joint structures such as ligaments or menisci.

MRI of cartilage requires close attention to imaging spatial resolution. To see early morphologic degenerative changes in cartilage, imaging with a resolution on the order of 0.2–0.4 mm is required [27]. The ultimate resolution achievable is governed by the SNR possible within a given imaging time and system. The use of high-field-strength magnets and dedicated phased-array or surface coils usually results in the best possible resolution in vivo. Eventually, a high-resolution imaging technique that provides morphologic and physiologic information together would be ideal in the evaluation of osteoarthritis. Given the current techniques, the combination of a high-resolution morphologic imaging sequence with a sequence for matrix evaluation will likely be the most useful.

2D Fast Spin-Echo Imaging

Currently, imaging of the musculoskeletal system with MRI is often limited to 2D multislice acquisitions acquired in multiple planes. These acquisitions are commonly performed with turbo or fast spin-echo (FSE) methods. These methods provide excellent SNR and contrast between tissues of interest, but the inherently anisotropic voxels in these 2D acquisitions require that multiple planes of data be acquired to minimize partial volume artifacts. For example, a typical sagittal image may have a 0.3- to 0.6-mm in-plane resolution but a slice thickness of 2–5 mm. FSE techniques show excellent results in the detection of cartilage lesions [28] (Figs. 1 and 2). These methods provide excellent depiction of structures in the imaging plane, but evaluation of oblique or small structures across multiple slices can be challenging. For these reasons, 3D acquisitions with thin sections are appealing.

3D Gradient-Echo Techniques

Traditional 3D gradient-echo methods have the potential to acquire data with more isotropic voxel sizes, but suffer from a lack of contrast compared with spin-echo approaches. High accuracy for cartilage lesions has been shown with 3D SPGR imaging [22,26,29]. There are two main disadvantages to this approach: lack of reliable contrast between cartilage and fluid that outlines surface defects and long imaging times, up to 8 minutes. In addition, SPGR imaging uses gradient and radiofrequency spoiling to reduce artifacts and achieve near T1-weighting, but this reduces the overall signal compared with steady-state techniques. Finally, 3D gradient-echo methods are less useful for the diagnosis of ligament or meniscal tears than spin-echo techniques. Despite these limitations, 3D SPGR imaging is considered the standard for morphologic imaging of cartilage [30,31].

The SPGR and gradient-echo techniques can be combined with water–fat separation methods such as iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL) fat–water separation to produce excellent quality water and fat images with high resolution, up to $0.3 \times 0.3 \times 1.0$ mm. The SPGR method suppresses signal from joint fluid, whereas the gradient-echo method accentuates it (Figs. 3 and 4). Compared with balanced steady-state free precession (SSFP), which is described later in greater detail, these methods are not only less SNR efficient, but also less sensitive to magnetic field inhomogeneity [32–34]. Therefore, an ideal 3D cartilage imaging sequence that provides an optimal combination of resolution, SNR efficiency, and minimal artifacts has yet to be established. However, a number of newer techniques have been established that improve cartilage imaging.

Dual-echo steady-state (DESS) imaging has proven useful for the evaluation of cartilage morphology [35–38]. This technique acquires two or more gradient echoes separated by a refocusing pulse and then combines both echoes into the image. This results in an image with higher T2*-weighting that has bright signal from both cartilage and synovial fluid.

New MRI Methods

Driven Equilibrium Fourier Transform Imaging

Driven equilibrium Fourier transform (DEFT) imaging has been used in the past as a method of signal enhancement in spectroscopy [39]. The sequence uses a 90° pulse to return magnetization to the z-axis, increasing signal from tissues with long T1 relaxation times such as synovial fluid. Unlike conventional T1- or T2-weighted MRI, the contrast in DEFT imaging is dependent on the ratio of the T1/T2 of a given tissue. For musculoskeletal imaging, the DEFT sequence produces contrast by enhancing the signal from synovial fluid rather than attenuating the signal from cartilage as in T2-weighted sequences. This results in bright synovial fluid at short TRs. At short TRs, DEFT shows greater cartilage-to-fluid contrast than SPGR, proton density FSE, or T2-weighted FSE [40]. DEFT imaging has been combined with a 3D echo-

planar readout to make it an efficient 3D cartilage imaging technique. Unlike with T2-weighted FSE, cartilage signal is preserved due to the short TE with the DEFT sequence. A high-resolution 3D data set of the entire knee using a 512×192 matrix, 14-cm field of view, and 3-mm slices can be acquired in about 6 minutes. Initial studies of cartilage morphology have been performed using DEFT imaging [41,42], but this technique has not been conclusively proven to be superior to 2D approaches. A sequence similar to DEFT that has been used in musculoskeletal imaging is FSE with driven equilibrium pulses, which is referred to as “DRIVE” [43].

Balanced SSFP Imaging

Balanced SSFP (bSSFP) MRI is an efficient, high-signal method for obtaining 3D MR images [44]. Depending on the manufacturer of the MRI scanner, this method is also called true fast imaging with steady-state precession (trueFISP, Siemens Healthcare), fast imaging employing steady-state acquisition (FIESTA, GE Healthcare), or balanced fast-field echo imaging (Philips Healthcare) [45]. With recent advances in MR gradient hardware, the bSSFP sequence can be used without being affected by banding or off-resonance artifacts that were previously a problem with this method. However, banding artifacts due to off-resonance are still an issue as TR increases or at 3 T. Hence, TR is usually kept below 10 milliseconds with these techniques, which limits overall image resolution. Multiple-acquisition bSSFP can be used to achieve higher resolution [46] at the cost of additional scanning time.

Many methods have been proposed to provide fat suppression with bSSFP imaging. If the TR is sufficiently short and the magnetic field is homogeneous, conventional fat suppression or water excitation pulses can be used [47]. Linear combinations of bSSFP [48] and fluctuating equilibrium MRI (FEMR) [49] use the frequency difference between fat and water and multiple acquisitions to separate fat and water. Intermittent fat suppression [50] uses transient fat-saturation pulses to suppress lipid signal. The IDEAL method (Fig. 5) uses multiple acquisitions to separate fat and water, but does not depend on the fat–water frequency difference to constrain the TR [51]. Rapid separation of water and fat can be achieved with phase detection [52].

Several studies have shown the utility of the bSSFP sequence for imaging articular cartilage [49,53–55]. Because of the bright synovial fluid and 3D nature of the acquisition, bSSFP may also be useful for imaging internal derangements of other structures including ligaments and menisci [56].

Vastly Interpolated Projection Reconstruction Imaging

Imaging of the knee with a combination of a 3D radial k-space acquisition and bSSFP has several advantages. Three-dimensional radial acquisitions are often undersampled in sparse, high-contrast imaging environments such as contrast-enhanced MR angiography (CE-MRA) to decrease imaging time. Vastly interpolated projection reconstruction (VIPR), first developed for time-resolved CE-MRA [57], was later adapted for bSSFP imaging of the musculoskeletal system. The radial acquisition allows a very efficient k-space trajectory that collects two radial lines for each TR without wasting time on frequency dephasing and rephasing gradients. One radial line begins at the k-space origin while the other is acquired along a different return path to the origin, allowing acquisition to occur during nearly the entire TR. The optimal TR needed for the most efficient implementation of linear combinations of bSSFP to do fat–water separation at 1.5 T (2.4 milliseconds) can be met while still having time for adequate spatial encoding.

Application of VIPR to the knee provides isotropic 0.5- to 0.7-mm 3D imaging that allows reformations in arbitrary planes. Because this method is based on SSFP, joint fluid is bright,

providing excellent contrast for diagnosis of meniscal tears, ligament injuries, and cartilage damage [58]. Contrast between the cartilage and bone is generated by separating fat and water with linear combinations of SSFP, as shown in Figure 6. Scanning time for the isotropic acquisition is only 5 minutes. An alternative single-pass method separates fat and water by exploiting the different phase progressions of fat and water spins between the two echoes acquired each TR. At 3 T, fat-and-water separation is achieved using an alternative fat stop band with a TR of 3.6 milliseconds. Here, the multiple-echo acquisition allows the removal of the unwanted passband between the water and fat resonance frequencies at the longer TR.

3D FSE Imaging

Two-dimensional FSE is a powerful clinical tool, but this method suffers from anisotropic voxels, slice gaps, and partial volume effects. Three-dimensional acquisitions with FSE were applied in brain imaging several years ago [59]. Three-dimensional FSE, with flip angle modulation to reduce blurring and parallel imaging to reduce imaging time, has made isotropic imaging with spinecho contrast a clinical reality [60,61]. Preliminary studies using 3D FSE with isotropic resolution in the knee [62] and ankle [63] have been published. Images from this method show isotropic resolution with the ability to obtain high-quality multiplanar reformations (Fig. 7). A recent study in more than 100 patients with arthroscopic correlation showed that 3D FSE was equal to a combination of multiple planes of 2D FSE in the diagnosis of ligament, menisci, and cartilage defects [64].

High-Field MRI

Several centers currently have 7-T human MRI systems available. Although these systems suffer from problems of radiofrequency penetration and high power deposition, they have a considerable SNR advantage over lower-field-strength systems. These systems are able to reach a higher resolution in a shorter period of time and may be useful for showing cartilage ultrastructure. Figure 8 shows a representative data set at 7 T from a healthy volunteer using a SPGR acquisition. This example shows that it is possible to acquire high-resolution ($0.3 \times 0.4 \times 1.5$ mm) morphologic data with excellent SNR in as little as 3 minutes.

Cartilage Thickness and Volume Mapping

Measurement of cartilage thickness and volume can be useful in tracking the progression of osteoarthritis [65]. Multicenter studies such as the Osteoarthritis Initiative use high-resolution 3D imaging of cartilage followed by manual or semiautomated segmentation of the images [35]. These data can show changes in cartilage thickness in high risk populations in as little as 1 year [66]. Figure 9 shows 3D reconstructions of knee cartilage from MRI data.

Physiologic MRI of Cartilage

Articular cartilage is approximately 70% water by weight. The remaining tissue consists predominately of type 2 collagen fibers and proteoglycans. The proteoglycans have GAG side chains with abundant negatively charged carboxylate and sulfate groups. Therefore, mobile ions such as sodium (Na^+) or charged gadolinium MRI contrast agents such as gadopentetate dimeglumine²⁻ distribute in cartilage in relation to the proteoglycan concentration. The collagen fibers have an ordered structure, making the water associated with them exhibit both magnetization transfer and magic angle effects. Physiologic MRI of articular cartilage takes advantage of these characteristics to explore the collagen and proteoglycan matrices for pathology. Although the methods described here can be performed at 1.5 T, all of them benefit from the additional SNR available on 3-T systems.

Relaxation Time Mapping

T2 Relaxation Time Mapping

MRI is characterized by excitation of water molecules and relaxation of the molecules back to an equilibrium state. The exponential time constants describing this relaxation are referred to as T1 and T2 relaxation times and are constant for a given tissue at a given MR field strength. Changes in these relaxation times can be due to tissue pathology or the introduction of a contrast agent.

The T2 relaxation time of articular cartilage is a function of both the water content and collagen ultrastructure of the tissue. Measurement of the spatial distribution of the T2 relaxation time may reveal areas of increased or decreased water content that correlate with cartilage damage. To measure the T2 relaxation time with a high degree of accuracy, attention must be taken with selecting the MR technique [67]. Typically, a multiecho spin-echo technique is used and signal levels are fitted to one or more decaying exponentials, depending on whether more than one distribution of T2 is thought to be within the sample [68]. However, for TEs used in conventional MRI, a single exponential fit is adequate. An image of the T2 relaxation time is then generated with either a color or a gray-scale map representing the relaxation time as shown in Figure 10.

Several investigators have measured the spatial distribution of T2 relaxation times within articular cartilage [69,70]. Aging appears to be associated with an increase in T2 relaxation times in the transitional zone [71]. Relaxation time measurements have also been shown to be anisotropic with respect to orientation in the main magnetic field [72–74]. Focal increases in T2 relaxation times within cartilage have been associated with matrix damage, particularly loss of collagen integrity. Studies on T2 relaxation times documenting the effects of age [75], sex [76], and activity [77] have also been published.

Contrast-Enhanced Imaging

One of the most common MRI contrast agents, Magnevist (Bayer HealthCare), or gadopentetate dimeglumine²⁻, has a negative charge. After IV injection of gadopentetate dimeglumine²⁻, it penetrates cartilage and concentrates where the cartilage GAG content is relatively low. Subsequent T1 imaging, which is reflective of gadopentetate dimeglumine²⁻ concentration, therefore yields an image depicting GAG distribution. This technique is referred to as delayed gadolinium-enhanced MRI of cartilage (dGEMRIC), with the “delay” referring to the time needed for the gadopentetate dimeglumine²⁻ to penetrate the cartilage tissue [78, 79]. A T1 map of the cartilage allows assessment of GAG content, with lower values corresponding to areas of GAG depletion.

A number of clinical cross-sectional studies on specified populations have provided interesting observations. For example, investigators have reported that individuals who exercise on a regular basis have higher dGEMRIC indices, denoting higher GAG, than those who are sedentary [80]. In a relatively large study of patients with hip dysplasia, measures of the severity of dysplasia—that is, the radiographically determined lateral center edge angle—and of pain both correlated with the dGEMRIC index, but not with the standard radiologic parameter of joint space narrowing [81]. In another study, lesions in patients with osteoarthritis were more apparent with the dGEMRIC technique relative to standard MRI scans [82]. A recent study relevant to osteoarthritis showed that dGEMRIC findings correlated with Kellgren–Lawrence radiographic grading of osteoarthritis [83]. Investigators have also studied the effects of gadolinium on measurement of T2 relaxation times [84,85].

Physiologic methods, such as dGEMRIC and T2 mapping, can be time-consuming and difficult to perform on a routine basis. The dGEMRIC technique is often performed using a 3D SPGR

approach with a variable flip angle as shown in Figure 11 [83, 86]. SSFP methods also show promise in improving the speed and SNR of T1 and T2 relaxation time measurements [87, 88]. Newbould et al. [89] recently developed an inversion recovery method of acquiring proton density, T1, and T2 maps using the SSFP sequence to image articular cartilage. Aside from generating quantitative T1, T2, and proton density maps, bSSFP images are also available for radiologic review. Quantitative techniques such as dGEMRIC and bSSFP may better elucidate physiologic changes in musculoskeletal imaging.

T1rho Imaging

A promising technique for evaluating cartilage is T1rho imaging, or relaxation of spins under the influence of a radiofrequency field [90,91]. This technique may be sensitive to early proteoglycan depletion [92–94]. In typical T1rho imaging, magnetization is tipped into the transverse plane and is “spin-locked” by a constant radiofrequency field. Recent advances in T1rho imaging have led to rapid Cartesian acquisition strategies for 3 T [95,96]. An example of a T1rho map from the tibia of a healthy volunteer is shown in Figure 12.

Sodium MRI

Atoms with an odd number of protons or neutrons possess a net nuclear spin and therefore exhibit the MR phenomenon. Sodium-23 (^{23}Na) is an example of a nucleus other than ^1H that is useful in cartilage imaging. The Larmor frequency of ^{23}Na is 11.262 MHz/T compared with ^1H at 42.575 MHz/T, so at 1.5 T the resonant frequency of ^{23}Na is 16.9 MHz, whereas it is 63 MHz for ^1H . The concentration of ^{23}Na in normal human cartilage is approximately 320 mM with T2 relaxation times between 2 and 10 milliseconds [97]. The combination of lower resonant frequency, lower concentration, and shorter T2 relaxation times than ^1H make in vivo imaging of ^{23}Na challenging. Sodium imaging requires the use of special transmit-and-receive coils as well as relatively long imaging times to achieve adequate SNR.

Sodium MRI has shown some promising results in the imaging of articular cartilage based on its ability to depict regions of proteoglycan depletion [98]. Sodium-23 atoms are associated with the high fixed-charge density present in proteoglycan sulfate and carboxylate groups. Some spatial variation in ^{23}Na concentration is present within normal cartilage [97]. Because of the short T2 relaxation times of sodium, imaging is often performed with a non-Cartesian trajectory [99]. Figure 13 shows an example of a sodium image of the entire knee of a healthy volunteer obtained with a 3D cones technique at 3 T [100]. High sodium concentration is seen throughout the normal cartilage. In cartilage samples, sodium imaging has been shown to be sensitive to small changes in proteoglycan concentration [101,102]. This method shows promise to be sensitive to decreases in proteoglycan concentration that occur in early osteoarthritis. Triple-quantum-filtered imaging of sodium in cartilage, which may be even more sensitive to early changes than sodium imaging, is also possible [103].

Discussion and Conclusions

MRI provides a powerful tool for the imaging and understanding of cartilage. Improvements have been made in morphologic imaging of cartilage in terms of contrast, resolution, and acquisition time. These improvements allow detailed maps of the cartilage surface to be developed that can be used to quantify both thickness and volume. Much progress has been made in imaging cartilage physiology and detecting changes in proteoglycan content and collagen ultrastructure. A summary of the current imaging methods for cartilage imaging is shown in Table 2.

The choice of a particular protocol for imaging articular cartilage depends greatly on patient factors. For many patients with internal derangement, imaging with standard FSE and/or 3D

SPGR sequences may suffice. For patients being considered for surgical or pharmacologic therapy, however, a more detailed evaluation may be required. For example, fast morphologic imaging along with evaluation of cartilage physiology may allow noninvasive evaluation of cartilage implants at different time points.

Current musculoskeletal MRI protocols include multiple planes of 2D FSE and 3D gradient-echo images. The new morphologic methods presented here such as VIPR, 3D FSE and/or bSSFP achieve isotropic resolution for an entire joint in a single acquisition. The isotropic data can then be reformatted into standard or oblique planes as needed, with slice averaging to improve SNR. Combining these methods with fat–water separation methods such as IDEAL provides water, fat, and combined images. Overall, isotropic imaging with reformations could produce a considerable savings in overall protocol time.

Additional studies are required to validate these isotropic methods for their sensitivity to common joint pathology such as meniscal tears. The fundamental trade-off between image resolution and SNR still limits our ability to image in vivo with extremely high resolution. Patient motion due to long imaging times may ultimately limit the resolution achievable at 1.5 or 3 T, so higher-field-strength systems may be required. Isotropic acquisitions could improve patient throughput or allow detailed studies of cartilage physiology with methods such as T2 mapping, T1rho mapping, or sodium MRI. A combination of fast high-resolution morphologic imaging methods with newer physiologic techniques could expand the sensitivity of MRI to early cartilage degeneration. Ideally, the combination of these techniques will lead to an MRI examination for cartilage that is brief and well tolerated but that provides important morphologic and physiologic data.

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Fig. 1. Multiple planes of fast spin-echo (FSE) images show cartilage damage in 54-year-old man. **A**, Coronal T1-weighted FSE image shows full-thickness cartilage loss over medial femoral condyle (*arrow*). **B**, Coronal T2-weighted FSE image obtained with fat saturation shows cartilage loss at same location (*arrow*), with overlying bone marrow edema. **C**, Sagittal intermediate-weighted FSE image shows full-thickness loss of cartilage (*arrow*). **D**, Sagittal T2-weighted FSE image with fat saturation shows full-thickness cartilage loss (*arrow*).

Fig. 2.

Axial fast spin-echo (FSE) images with fat saturation of cartilage damage.

A, Axial intermediate-weighted FSE image shows grade 1 (increased signal) cartilage damage in lateral facet (*arrow*) in 35-year-old woman.

B, Axial intermediate-weighted FSE image shows grade 2 cartilage loss of less than 50% (*arrow*) over medial facet in 34-year-old man.

C, Image shows more extensive grade 3 cartilage loss greater than 50% over medial facet (*long arrow*) and deep fissure in lateral facet (*short arrow*) in 54-year-old man.

D, Image shows full-thickness defect (grade 4) over trochlea with delamination and fragment (*arrow*) in 27-year-old man.

Fig. 3. Sagittal 3D gradient-echo images from ankle of healthy 35-year-old male volunteer at 3 T. **A**, Spoiled gradient-recalled echo image shows dark synovial fluid. **B**, Gradient-recalled echo image obtained without radiofrequency spoiling results in T1/T2 contrast and bright synovial fluid (*arrow*).

Fig. 4. Sagittal 3D gradient-echo images from knee of healthy 35-year-old male volunteer at 3 T. Fat–water separation was done using iterative decomposition of water and fat using echo asymmetry and least-squares estimation (IDEAL).
A, IDEAL spoiled gradient-recalled image shows dark synovial fluid (*arrow*).
B, IDEAL gradient-recalled echo (GRE) image shows bright synovial fluid (*arrow*) (flip angle = 14°).
C, IDEAL GRE image shows bright synovial fluid and darker cartilage (*arrow*) (flip angle = 25°).

Fig. 5. Balanced steady-state free precession (bSSFP) images of knee of healthy 32-year-old male volunteer acquired using iterative decomposition of water and fat with echo asymmetry and least-squares estimation (IDEAL) bSSFP. **A and B,** Water image (**A**) and fat image (**B**). Note that joint fluid is bright in **A** using this bSSFP technique. (Reprinted with permission of Radiological Society of North America from [55])

Fig. 6.

Vastly interpolated projection reconstruction (VIPR) balanced steady-state free precession (SSFP) imaging of knee in 27-year-old woman at 3 T. This SSFP-based technique produces 0.4-mm isotropic resolution across knee, allowing reformations in any imaging plane. Scanning time was only 5 minutes. (Courtesy of W. Block, University of Wisconsin, Madison)

A, Coronal image, 2-mm section thickness.

B, Sagittal reformation, 2-mm section thickness.

C, Axial reformation, 2-mm section thickness.

D, VIPR k-space trajectory.

Fig. 7.

Three-dimensional fast spin-echo imaging using flip angle modulation and parallel imaging in 54-year-old man. This acquisition was done at 3 T with imaging time of 5 minutes and isotropic 0.6-mm resolution. (Courtesy of S. Majumdar, University of California, San Francisco)

A, Coronal image, 2-mm section thickness.

B, Sagittal reformation, 2-mm section thickness.

C, Axial reformation, 2-mm section thickness.

Fig. 8. Sagittal images of healthy 24-year-old female volunteer at 7 T using 3D spoiled gradient-echo method. Resolution was $0.3 \times 0.4 \times 1.5$ mm.
A, Image obtained with no parallel imaging. Scanning time was 6 minutes 18 seconds.
B, Factor of 2 parallel acceleration. Scanning time was 3 minutes.

Fig. 9.

Three-dimensional volume renderings of cartilage in 45-year-old man from segmented high-resolution MRI. (Courtesy of J. Hohe and F. Eckstein, Chondrometrics GmbH, Germany)

A, Posterior view. Medial tibial cartilage is labeled blue; lateral tibial cartilage, green; medial femoral condyle, orange; and lateral femoral cartilage, purple.

B, Lateral view. Lateral tibial cartilage is labeled green; femoral trochlea, turquoise; weight-bearing part of lateral femoral condyle, red; and posterior part of lateral femoral condyle, violet.

Fig. 10. Color map of T2 relaxation time in tibial cartilage in healthy 44-year-old male volunteer at 3 T. T2 mapping is sensitive to status of collagen matrix in articular cartilage. This map was acquired with Cartesian 2D fast spin-echo technique with imaging time of 10 minutes.

Fig. 11. Variable nutation angle spoiled gradient-echo (SPGR) T1 maps of cartilage after IV contrast administration in 37-year-old man.
A and B, These maps were acquired at 3 T with 0.4-mm in-plane resolution and 2-mm section thickness. Eight flip angles were used to create T1 maps. Delayed gadolinium-enhanced MR image of cartilage with T1 maps such as these is sensitive to proteoglycan status in cartilage.

Fig. 12.

Color map of T1rho relaxation time in tibial cartilage in healthy 44-year-old male volunteer at 3 T. T1rho mapping is sensitive to status of proteoglycan in articular cartilage. This map was acquired with Cartesian 2D spoiled gradient-echo technique with imaging time of 10 minutes.

Fig. 13.

Three-dimensional cones acquisition of sodium MRI in healthy 28-year-old female volunteer at 3 T. Sodium is marker for proteoglycan in articular cartilage. This image was obtained with custom dual-tuned knee coil and registered to proton spoiled gradient-echo image. Resolution was $1.25 \times 1.25 \times 4$ mm, and imaging time was 21 minutes. Test tube is in place for sodium quantification of sodium signal.

TABLE 1

Modified Noyes Score

Modified Noyes Score	MRI Finding
0	Normal
1	Signal change (increased T2)
2	Partial-thickness defect < 50%
3	Partial-thickness defect $\geq$ 50%
4	Full-thickness defect

TABLE 2

Current Imaging Methods for Cartilage Imaging

Method	Features
2D fast spin-echo	Standard for clinical imaging Blurring on proton density images Slice gaps Anisotropic resolution
3D gradient-echo	Standard for 3D cartilage morphology T2/T1 weighting Spoiling reduces fluid signal
3D bSSFP including VIPR	T2/T1 weighting Bright fluid Highly efficient Banding artifacts at long TRs
3D fast spin-echo	Similar to traditional fast spin-echo contrast Long echo trains, potential blurring Parallel imaging Isotropic resolution with reformats
T2 mapping	Sensitive to cartilage–collagen matrix Magic angle dependence
T1rho mapping	Sensitive to cartilage proteoglycan High radiofrequency power deposition
dGEMRIC	Sensitive to cartilage proteoglycan Requires double-dose IV contrast injection 90-min delay before imaging
Sodium MRI	Dedicated coil hardware Direct measure of cartilage proteoglycan High-field MRI required

Note—bSSFP = balanced steady-state free precession, VIPR = vastly interpolated projection reconstruction,dGEMRIC = delayed gadolinium-enhanced MRI of cartilage.