Imaging of short and ultrashort $T_2$ and $T_2^*$ tissues using clinical MRI systems

There are now a variety of new techniques available to detect signal from tissues with short or ultrashort $T_2$s and $T_2^*$s. There are also many methods of developing image contrast between tissues and fluids in the short $T_2$ or $T_2^*$ range, which can provide visualization of anatomy that has not previously been seen. Particular methods have been developed to target susceptibility effects, and allow accurate quantitation by compensating for the anatomical distortion produced by these effects. Specific methods have been developed to image the effects of magnetic iron oxide particles with positive contrast and to correct for the loss of signal and image distortion near to metal caused by gross susceptibility effects. These methods are likely to provide interesting options and increase the range of applications of MRI.

**KEYWORDS**: short $T_2$ tissue components, susceptibility, ultrashort echo time

During the first year of clinical MRI, only steady state free precession, $T_1$-weighted and proton density-weighted clinical images were available [1–3]. Heavily $T_2$-weighted spin echo (SE) sequences arrived suddenly in early 1982 and transformed the practice of magnetic resonance (MR) [4–6]. Images obtained with these sequences detected intermediate or long $T_1$ relaxation components in tissue. Even with the subsequent development of new classes of sequences, such as fast SE, clinical diffusion-weighted imaging and fluid attenuated inversion recovery, the detection of signal from intermediate and long $T_1$ relaxation components remains the dominant form of MRI for the diagnosis of parenchymal disease in the brain and much of the rest of the body.

However, even when clinical MRI began, very short mean $T_2$ relaxation components were recognized in the cortical bone by Smith et al. [7] and Edelstein et al. [8]. This tissue showed no MR signal. The lack of signal was useful in providing a low-signal background against which abnormalities in cortical bone with sufficiently long mean $T_2$s to result in detectable signal could be recognized, but the absence of signal from normal cortical bone meant that there was no possibility of measuring normal values of mobile proton density ($\rho_m$), $T_1$ or $T_2$. Nor was it possible to study normal perfusion, and there was no opportunity for active contrast manipulation, little or no distinction between adjacent short $T_2$ tissues and no means of visualizing normal contrast enhancement. As a result, the study of cortical bone and other MR ‘invisible’ short $T_2$ tissues, such as tendons, ligaments and menisci, has been far more limited than that of tissues and organs, such as brain, liver and muscle, where tissue mean $T_2$s are longer and MR signal from them is readily detectable with conventional clinical sequences. However, even these longer $T_2$ tissues contain significant proportions (e.g., 5–30%) of invisible or undetectable short $T_2$ relaxation components when they are imaged with conventional approaches.

To image short or ultrashort $T_2$ tissues that produce no detectable signal with conventional sequences, indirect methods have been used in which signal is obtained from surrounding or associated longer $T_2$ tissue. When the low- or zero-signal tissue is surrounded by longer $T_2$ tissue, signal from this tissue can be used to define the boundaries of the zero-signal tissue. It is also possible to characterize some short $T_2$ tissues by observing the impact that their difference in susceptibility from that of the surrounding longer $T_2$ tissues has on the signal obtained from the longer $T_2$ tissue. For example, some features of trabecular bone can be inferred by the effect this tissue has on the MR signal of adjacent red or yellow bone marrow [9]. A third indirect method is possible when short and long $T_2$ relaxation components are associated, and undergo magnetization exchange. The effect of saturation of the invisible short $T_2$ components on this exchange can be observed on the detectable longer $T_2$ components [10] and, thus, inferences can be made regarding the short $T_2$ tissue and/or the exchange between the shorter and longer $T_2$ components.
An alternative to using conventional sequences to study short \( T_2 \) tissues in this way is to employ methods that directly detect signal from them. These usually involve the use of short echo time (TE) or ultrashort TE (UTE) sequences to detect MR signals before they have decayed to zero. There are now a variety of sequences of this type available in the clinical domain.

While \( T_2 \) is a property of tissue that reflects dipolar and other nuclear interactions, frequently the effects seen with MRI are described more accurately by the observed \( T_2^* \) or \( T_1^* \). This includes effects such as intravoxel dephasing due to \( B_0 \) field inhomogeneity, tissue susceptibility differences and chemical shift. Tissue susceptibility effects reflect the fact that solid tissues, such as bone, are generally more diamagnetic than soft tissues, and that some tissues and fluids may be paramagnetic. The effects of some of these differences can be partly or almost wholly reversed by the use of SE sequences.

In some situations, \( T_2^* \) effects may dominate and it is useful to recognize several different approaches to imaging short \( T_2^*/T_1^* \) components in relation to underlying susceptibility differences:

- The first approach essentially sees the problem as imaging of short or ultrashort \( T_2 \) components and the basic approach is to use a short TE or UTE to acquire and encode MR signals before they decay to a low level. This may be appropriate in situations where there are only minor susceptibility differences present.
- The second is susceptibility-weighted imaging, where magnitude and phase data are used to recognize a loss of signal from tissue caused by susceptibility effects. It can be direct and/or indirect (where \( T_2^* \) become too short to detect) and is qualitative.
- Quantitative susceptibility imaging is the third approach. This technique recognizes the fact that susceptibility differences affect the spatial encoding of MR signals and endeavors to correct this and to calculate values of \( T_2^* \) that accurately reflect \( T_2 \) and susceptibility effects.
- Positive-contrast and white-marker imaging techniques address the specific problem of imaging the effects of magnetic iron oxide particles (MIOPs), which shorten \( T_1 \) and produce local disturbances of the magnetic field. The aim is to detect the presence of particles with a positive signal and at least, in part, address the problem of field distortion to achieve credible recognition and quantification of the concentration of MIOPs.
- The fifth group of techniques is targeted at imaging in the presence of metal. Metals can produce very large susceptibility effects with loss of signal due to \( T_2^* \) effects and gross image distortion. The primary objective in this situation is to deal with the image distortion and restore image integrity to a sufficient degree for the images to be clinically useful.

There is some overlap between these approaches. In some situations it may be appropriate to ignore the effects of susceptibility differences in producing image distortion and regard the problem as one of detecting short \( T_2 \) signals, whereas in other situations image distortion due to susceptibility is the primary problem that needs to be addressed. Over the previous year, there has been considerable interest in these approaches with solutions now appearing to some problems that were previously intractable for many years.

**Tissue properties**

The tissues of the human body can be divided into those that are visible, in the sense that they provide detectable signal with clinical MR systems, and those that are ‘invisible’ because their mean \( T_2^* \)s or \( T_1^* \)s are too short to provide a detectable signal. All tissues have multicomponent \( T_2^* \)s. This means that they contain a mixture of short and long \( T_2 \) components. The invisible tissues have a majority of short \( T_2 \) components and a minority of long \( T_2 \) components. The latter components typically do not provide enough signal to be detectable in comparison to image noise levels. The invisible tissues of the body, such as the brain, liver and muscle, have a majority of long \( T_2 \) components, which produce the signals seen with conventional techniques. They also have a minority of short \( T_2 \) components that do not contribute significantly to the detectable signal.

There is no agreement as to what constitutes a short TE and what is an UTE, and there is an argument regarding how TE should be measured for tissues with short \( T_2^* \)s [11], but for simplicity, a short TE is taken to be less than 10 ms and a UTE less than 1 ms. It is also possible to define short \( T_2^*/T_1^* \) as less than 10 ms and ultrashort as less than 1 ms. This reflects the fact that with older systems and SE sequences,
tissues with a $T_1$ or $T_2^*$ less than 10 ms produced little or no signal and were invisible. With more recent systems and gradient-echo sequences the cut-off is closer to 1 ms.

Within the invisible group of tissues (mean $T_2 < 10$ ms) it is possible to differentiate a first group, including tendons, ligaments and menisci, with short mean $T_2$s of approximately 1–10 ms, a second group, including cortical bone and dentine, with ultrashort mean $T_2$s of 0.1–1 ms. There is also a third group, including dental enamel, protons in membranes, large molecules and crystalline bone, with a mean $T_2$ of less than 0.1 ms.

An important factor in this context is the magic-angle effect [12,13] since it can greatly increase the $T_1^*$ of short $T_2$ tissues, such as tendons, ligaments and menisci. When the orientation of tissues that contain highly ordered collagen is changed, their $T_1^*$ varies from a minimum at $\theta = 0^\circ$, where dipolar interactions are greatest, to a maximum where $3\cos^2\theta - 1 = 0$ and $\theta = 55^\circ$. $\theta$ is the orientation of the fibers to $B_0$. The increase can be large, for example, from 0.6 to 21 ms [12] or from 7 to 23 ms [13] in the Achilles tendon.

A recently described phenomenon is directional susceptibility in tendons, whereby their bulk magnetic susceptibility varies with orientation to $B_0$, with signals at the water end of the proton spectrum when fibers are parallel to $B_0$ and at the fat end of the spectrum (lower frequency) when fibers are perpendicular to $B_0$ [14]. The difference is relatively large (of the order of three parts per million [ppm]).

The $\rho_m$ of tissues also varies markedly, with bone having a $\rho_m$ of 15–20% and semi-solid tissues, such as tendons and ligaments, having values of 60–70%. $\rho_m$ is generally a more important factor in generating contrast with short $T_2$ tissues than it is with longer $T_2$ tissues. The low $\rho_m$ for bone places a limit on the maximum signal that can be obtained from it.

The mean $T_2$s of some tissues with a majority of short $T_2$ components are short, with cortical bone having a particularly short $T_2$, in fact, less than that of fat [15]. The relative differences in mean $T_2$ or $T_2^*$ between normal and abnormal tissue are generally much greater than those in mean $T_1$.

Relative to air, soft tissues generally show a susceptibility difference of approximately -9 ppm, and bone and calcified tissue of approximately -11 ppm. By comparison, the principal peak of fat resonates at approximately -12 ppm. Paramagnetic materials may show small positive chemical shifts and superparamagnetic materials show greater positive shifts again. Metals including, for example, titanium and some types of stainless steel, may show large positive shifts of 10–100s ppm (or more). These changes in field may be considerably greater than those used by applied machine gradients to encode MR signals and may therefore cause image distortions.

### Acquisition methods for short $T_2/T_2^*$ components of tissue

Some of the techniques currently being used to detect signal directly from tissues on clinical systems have been used in material science and tissue studies using small-bore high-field spectrometers for many years. Methods now in use on clinical systems are summarized in Table 1. The prototype sequence for imaging short $T_2$ tissues is single-point imaging where a single point in k-space is acquired with an UTE. This is typically used with 3D phase encoding, which unfortunately makes the technique very time consuming [16].

It is possible to acquire several points at a time, which makes the sequences more time efficient but results in longer TEs for the additional points [17]. There are also free induction decay-based techniques where a radial line of k-space is acquired from the center out [18]. This can be coupled with long $T_2$ water- and fat-suppression to selectively image short $T_2$ components [19]. Other trajectories in k-space are possible, including a stack of spirals [20].

### Table 1. Short and ultrashort echo time imaging techniques.

<table>
<thead>
<tr>
<th>Technique</th>
<th>Radiofrequency pulses and gradient</th>
<th>k-space trajectory</th>
<th>Ref.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single point</td>
<td>Nonselective hard pulse with gradient applied</td>
<td>3D point by point</td>
<td>[16]</td>
</tr>
<tr>
<td>Multipoint</td>
<td>Hard pulse with gradient applied</td>
<td>3D partial lines, several points</td>
<td>[17]</td>
</tr>
<tr>
<td>Ultrashort echo time</td>
<td>2D two half pulses</td>
<td>Radial, from center out</td>
<td>[18]</td>
</tr>
<tr>
<td>Water- and fat-suppressed proton projection MRI</td>
<td>3D hard pulse with gradient applied during radiofrequency</td>
<td>Radial, from center out</td>
<td>[19]</td>
</tr>
<tr>
<td>Gradient echo</td>
<td>2D, 3D</td>
<td>Radial rephasing gradients</td>
<td></td>
</tr>
<tr>
<td>Cones, spiral, stack of spirals</td>
<td>3D</td>
<td>Spiral, from center out</td>
<td>[20,57]</td>
</tr>
<tr>
<td>SWIFT, SEA</td>
<td>3D radiofrequency subpulses</td>
<td>Radial, from center out</td>
<td>[21–24]</td>
</tr>
</tbody>
</table>

*FID: Free induction decay; SEA: Simultaneous excitation and acquisition; SWIFT: Sweep imaging with Fourier transform.*
A particularly innovative method of imaging short $T_2$ components is to divide the excitation pulse into subpulses and acquire data after each of these pulses. The acquired data need to be deconvolved with the excitation pulse, but the end result is a much more time efficient acquisition than with typical 3D acquisitions [21–24]. Other techniques that have only been used in the preclinical phase include methods in which radiofrequency absorption, rather than signal detection, is assessed [25]. The methods borrow from older forms of spectroscopy and electron spin resonance, where electronic $T_2$s are extremely short and may be of the order of a microsecond.

**Magnetization preparation, contrast mechanisms & signal-suppression techniques**

Traditional contrast mechanisms exploiting differences in $\rho_m$, chemical shift and other tissue properties can be used in ways that are well known from conventional imaging.

There are also numerous new contrast mechanisms, or old contrast mechanisms operating in new ways, that are of interest in imaging short/ultrashort $T_2/T_2^*$ components in tissue. Some of these are listed in Table 2. They are typically used in conjunction with the acquisition techniques detailed in the previous section. These provide a wide range of possible ways of effecting magnetization. For example, 90° and 180°, fat saturation and magnetization transfer pulses can all be used to suppress unwanted long $T_2$ signals and to produce $T_2$ contrast in the short $T_2$ range. There are also relatively new potential mechanisms involving double quantum filters [26] and a reduction in dipolar coupling [27,28]. These techniques are usually applied in conjunction with one of the acquisition methods described in the previous section.

**Susceptibility-weighted imaging**

Susceptibility-weighted imaging has been in use for a considerable time. It usually exploits reductions in $T_2^*$ to develop contrast, and imaging may utilize both magnitude and phase data [29,30]. The $T_2^*$ may be so short that this, in effect, becomes an indirect form of imaging utilizing the reduction in signal in adjacent longer $T_2$ components. The applicability of the technique and related methods can be expanded by utilizing forms of data collection with TEs or UTEs that can detect signal from very short $T_2^*$ components [31].

**Quantitative susceptibility imaging**

Quantitative methods of imaging susceptibility changes need to account for errors in spatial encoding, which may require solutions to a complex inverse problem [32,33]. To date, it has mainly been applied to brain imaging.

**Positive-contrast & white-marker imaging**

These forms of imaging have been used to describe the particular situation with MIOPs that may not only reduce $T_2$ and $T_2^*$, but also

<table>
<thead>
<tr>
<th>Mechanism</th>
<th>Effect</th>
<th>Ref.</th>
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<tbody>
<tr>
<td>90° pulse</td>
<td>Selective excitation of short $T_2/T_2^*$ components with or without subsequent long $T_2$ signal suppression</td>
<td></td>
</tr>
<tr>
<td>180° pulse</td>
<td>Selective excitation of short $T_2/T_2^*$ components and inversion of long $T_2$ components</td>
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<tr>
<td>180° pulse and nulling</td>
<td>Selective inversion of long $T_2/T_2^*$ components with nulling</td>
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<tr>
<td>Off-resonance saturation</td>
<td>Selective reduction of short $T_2$ components</td>
<td>[58]</td>
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<tr>
<td>Magnetization transfer</td>
<td>Selective reduction of short $T_2$ components with magnetization transfer to detectable $T_2$ components</td>
<td>[59]</td>
</tr>
<tr>
<td>Fat saturation</td>
<td>Selective reduction of fat and short $T_2/T_2^*$ water signals</td>
<td></td>
</tr>
<tr>
<td>Later image subtraction from first image</td>
<td>Selective reduction of long $T_2/T_2^*$ components</td>
<td></td>
</tr>
<tr>
<td>Susceptibility and spectral mapping</td>
<td>Direct mapping of field change and susceptibility differences</td>
<td>[34]</td>
</tr>
<tr>
<td>ultrashort echo time spectroscopic imaging</td>
<td>Selective imaging of protons with strong dipolar coupling</td>
<td>[26]</td>
</tr>
<tr>
<td>Double quantum filter</td>
<td>Comparison of spin echo and magic sandwich echo imaging</td>
<td>[27]</td>
</tr>
<tr>
<td>Dipolar imaging</td>
<td>$T_2$ in rotating frame</td>
<td></td>
</tr>
<tr>
<td>Gd contrast agents</td>
<td>Detectable short $T_2/T_2^*$ in tissues</td>
<td>[60]</td>
</tr>
<tr>
<td>Magnetic iron oxide particles</td>
<td>Reduction in detectable signal in short $T_2$ tissues</td>
<td>[61]</td>
</tr>
<tr>
<td>High velocity flow</td>
<td>Phase-shift due to flow can be specifically targeted</td>
<td>[62]</td>
</tr>
</tbody>
</table>

Gd: Gadolinium.
produce local field distortions. A variety of different methods are available. It is possible to select ‘actively’ only off-resonance spins. It is also possible to apply an additional gradient so that only the magnetization of spins in regions affected by MIOPs are refocused. The inhomogeneities from the particles induce echo shifts and these can be used to calculate and correct for the field distortion. The images reflect both tissue MIOP concentration and deviations of the local magnetic field produced by the particles [34–38].

Imaging in the presence of metal

When forms of metal are implanted in the body, an extreme situation may arise in which there is very marked \( T_2^* \) shortening, but the image distortion is so great that images of regions adjacent to the metal are uninterpretable. This has been a longstanding problem. In the past, a variety of solutions have been proposed, but these have had relatively little clinical impact. The recent development of multi-acquisition variable-resonance image combination (MAVRIC) [39] and slice encoding for metal artifact correction (SEMAC) [40] has resulted in a remarkable degree of restoration of images that are grossly degraded by a metallic artifact when imaged using conventional approaches. With MAVRIC, irradiation at a range of different off-resonance frequencies is used to detect signals whose resonant frequency has been shifted by metal. With SEMAC, phase encoding is used during slice selection to reallocate signals that are improperly located by the slice selection process. View angle tilting is also used with this technique to correct for errors with in-plane spatial encoding [41].

Imaging of boundaries involving short \( T_2 / T_2^* \) tissues

Structures of interest in the short \( T_2 \) range include thin layers such as those in entheses, periosteum and the deep layers of articular cartilage where there are short \( T_2 \) tissues, susceptibility effects between the soft (or semi-solid) tissues and bone, as well as partial volume effects between these tissues over curved surfaces. In this situation, high-resolution 3D-isotropic UTE imaging often has a distinct advantage since it can detect short \( T_2 / T_2^* \) signals as well as reduce the impact of susceptibility differences and partial volume effects. Imaging of ordered fibrous structures, such as tendons and ligaments, include some of the previously discussed issues but, in addition, a loss of contrast of the fiber structure or a blurred appearance may arise from obliquity of the fibers relative to the imaging slice. This Filler effect may simulate changes due to disease. There are also distinctive artifacts at boundaries from chemical shift effects, including those associated with radial acquisitions.

Clinical applications

There are now 2D and 3D UTE sequences available with imaging times of 5–6 min and clinically acceptable spatial resolution. In general the difficulty of acquiring short/ultrashort \( T_2 / T_2^* \) signals means that invisible tissues are imaged at lower spatial resolutions, but with signal levels and contrast that are not attainable with conventional techniques. A balance is necessary to obtain novel qualitative and/or quantitative information at spatial resolutions that show anatomical features with acceptable clarity.

- Cortical bone

Cortical bone can be imaged with high signal [15]. The \( T_2 \) is approximately 0.4 ms and \( T_1 \) 250–350 ms at 1.5 T, which is shorter than typical values for fat. The mobile proton density is approximately 15–20%. These data can be used both for quantitative [42] and qualitative studies (Figure 1).

- Tendons, ligaments & entheses

In conventional sequences, the signal from tendons, ligaments and entheses is very low or zero. Entheses are the attachment sites of tendons, ligaments and capsules to bone. They typically contain calcified and uncalcified fibrocartilage (which both have short \( T_2^* \)). These tissues have a major role in dispersing mechanical stress at the junction between flexible tendons or ligaments and rigid bone.

Figure 1. Ultrashort echo time MRI of the skull. The inner and outer tables of the skull are seen in a manner similar to x-ray CT with bone windows.
Tendons and ligaments contain endotenon and endoligament, which have longer $T_2$ than the fibrous components (although they are still in the short $T_2$ range) and less magic-angle effect. Uncalcified fibrocartilage has a longer $T_2$ than the tensile components of tendons as well as an increase in $T_2$ caused by the magic-angle effect, but this may be present over a wider range of angles reflecting the more dispersed arrangement of the fibers within it. Magic-angle effects may result in a longer $T_2$ adjacent to bone from fibers that change in direction, from parallel to the bone surface to perpendicular to it, as they insert into bone. Tendons and ligaments can readily be seen with UTE sequences and entheses have been studied in detail [43,44]. Off-resonance fat-suppression pulses reduce the signal from short $T_2$ fibers (which have a broad line width) more than endotenon or enthesis fibrocartilage (which have longer $T_2$ and narrower line widths) and this can be an effective contrast mechanism. Inversion pulses may be used to selectively invert and null enthesis fibrocartilage (exploiting its longer $T_2$), and so, visualize this tissue with high contrast. It is also possible to visualize oblique and transverse fibers in tendons using a combination of fat-suppressed UTE sequences to reduce short $T_2$ tissue water components and magic-angle imaging to lengthen the $T_2$ of the fibers at particular angles to $B_0$ (Figure 2).

Entheses are selectively involved in the sero-negative spondyloarthropathies, such as anklyosing spondylitis and psoriatic arthropathy. The differential diagnosis is of a loss or reduction in fascicular pattern and includes normal sesamoid fibrocartilage, partial volume effects with a loss of fascicular pattern due to the Filler effect, magic-angle effects and disease.

Menisci of the knee
The central region of the adult meniscus has no blood supply (the white zone) while the more peripheral region (the red zone) has a blood supply. Healing of tears in the white zone is generally unsatisfactory and the preferred surgical strategy is usually resection of the torn tissue. Suture and repair is more successful in the red zone. Distinction between the two zones has not previously been possible with MRI using conventional sequences, despite repeated attempts [45]. However, using UTE sequence and gadolinium-based contrast enhancement, the two zones can be distinguished [46] and provide a basis for surgical planning.

Anatomical descriptions of the meniscus include circumferential, radial, lamella, vertical and meshwork fiber groups. With conventional imaging, some radial fibers may be distinguishable from the majority of circumferential fibers, but with UTE and magic-angle imaging each of these fiber groups can be identified (Figure 3). It is also possible to distinguish the internal structure of the meniscus from that of the root ligaments, and the more central cartilaginous region from the more fibrous peripheral region of the meniscus.

The fiber structure provides a basis for understanding the biomechanics of the knee and the various patterns of tear in the meniscus. It also helps in distinguishing magic-angle effects within fiber groups from degenerative changes.

Temporomandibular joint disc
This demonstrated some of the characteristics of the meniscus of the knee. Fiber structure can be seen. Lamella, circumferential antero–posterior and superior–inferior fibers are identifiable.

Articular cartilage
Articular cartilage has a range of $T_2$s from approximately 1 to 30–40 ms from deep to superficial. With conventional imaging, the deep radial and calcified layers as well as the adjacent subchondral bone are invisible. With UTE imaging, signal is detectable from the deeper layers of cartilage, allowing more superficial cartilage and

Figure 2. Ultrashort echo time image of the Achilles tendon. Abnormal oblique fibers are seen within the tendon (arrows).
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subchondral bone to be distinguished [47]. This provides a basis for study of the junction between cartilage and bone, which may be of importance in the pathogenesis of osteoarthritis. Complex magic-angle effects are seen, owing to the fibrous architecture of articular cartilage.

In disease there may be both a loss of signal from the deep layer and an increased extent of the short $T_2$ associated with deep layers. There is electron microscope evidence of thinning of the deep layers in osteoarthritis but preservation in osteomalacia.

**Spine**

Imaging of the spine includes many visible tissues, therefore, to date, attention has focused on invisible structures such as entheses, the end plate of the disc, and short $T_2$ components in the intervertebral discs and red bone marrow. Fibrocartilage has also been demonstrated in the functional entheses of the transverse ligament of the atlas and the alar ligament. The dorsal capsule of the facet joints of the lumbar spine are also subject to cartilaginous metaplasia. Evidence of iron deposition can be seen in intervertebral discs in thalassemia [48].

**Brain**

There are significant short $T_2$ components in many tissues of the body with longer mean $T_2$'s, including the brain, liver and muscle. These components can be specifically detected using UTE and other acquisition methods coupled with techniques that suppress long $T_2$ signals [49,50].

**Liver**

The liver contains a relatively high proportion of short $T_2$ components. The $T_2^*$'s of these may be prolonged in fibrosis [51]. The fibrosis in this situation is often of a relatively open structure and includes free water.

**Pelvis**

Ultrashort echo time sequences have been applied to study the effects of cryosurgery in carcinoma of the prostate [52]. Freezing of tissues results in a substantial reduction in $T_2^*$. There are issues with measuring $T_2$ and $T_2^*$ in the correct range, characterizing different $T_2$ components (e.g., long and short) including their relative proportions and dealing with artifacts from various sources. Quantitation may be confounded by slice selection and eddy current problems, and by contamination of short $T_2$ components with long $T_2$ components that are present in higher concentration.

**Future perspective**

Imaging of short $T_2$ and $T_2^*$ components is a rapidly expanding area that has seen a convergence of methods targeted at tissues with short $T_2$ components, susceptibility effects, MIOP imaging and metal artifact control. The methods have improved on solid state imaging, spectroscopy (including continuous-wave methods), electron spin resonance and MR microscopy. The much lower technical performance of clinical systems compared with small-bore spectrometers is a major limitation, but innovative methods for overcoming this problem are now being developed.

The tissues of interest have mainly been in the musculoskeletal system but all tissues of the body have some short $T_2$ components and study of these may prove to be of diagnostic interest. Some techniques, such as imaging in the presence of metal, are likely to be useful in the clinical domain immediately, while others may require validation and comparative assessment to establish their role. Sodium and phosphorous studies may also be of interest [55,56]. Quantitative approaches may be particularly useful given the large fractional changes in $T_2$

Figure 3. Short echo time image of the meniscus. The lamellar layer, radial and vertical fibers are seen. Circumferential fibers are of low intensity.
and T2* that are frequently seen in disease. The techniques used for imaging often require high gradient performance with control of short-term eddy currents to a level not previously necessary in clinical MR systems. Despite these and other technical difficulties, application to the study of short T2 and T2* tissues appears likely to be an area of MRI of considerable importance in the near future.

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**Executive summary**

- New forms of data acquisition allow direct imaging of short mean T2 and/or T2* tissues.
- Imaging of the short T2 components in visible tissues may be important.
- New techniques for imaging the effects of susceptibility and quantifying these effects are likely to be important.
- Imaging in close proximity to metal is now possible.
- New options for the use of gadolinium chelates and magnetic iron oxide particles are now available.

**Bibliography**

Papers of special note have been highlighted as:

* of interest
** of considerable interest


**Description of advanced method of acquisition and data collection.**


**Good background reading and description of the continuous-wave method.**
Good general coverage of 


Key reference on metal artifact control.


Key reference on metal artifact control.


Clinical application of short T₂ measurements.


