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THE EVALUATION OF CHANGES IN THE KNEE MENISCUS IN VIVO AT 3T MRI SCANNER

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ABSTRACT

Noninvasive imagining of the knee meniscus without the use of the contrast agents is more difficult compared to articular cartilage. Despite the lower signal intensity of the knee meniscus, MRI is considered the best non-invasive imaging method. Thanks to the lower water content in the meniscus compared to the surrounding tissues, it can be distinguished from the environment, but the determination of the boundaries is more complicated than in articular cartilage. There are many studies dealing with the MR imaging of the loaded and also unloaded knee, but they have mainly observed quantitative and geometric changes (movement or deformation of tissue), not targeted qualitative changes in the extracellular matrix (ECM). These changes can be evaluated with T2 relaxation times, which are more sensitive to the interaction of water molecules and the concentration of macromolecules and structures of the ECM, especially in the interaction based on the content, orientation and anisotropy of collagen fibers. Fluid and tissues with the higher water content level have long relaxation time T2. In the healthy meniscus these times are shorter; the reason is a highly organized structure of collagen and lower content of proteoglycans. To quantitatively detect changes, it is necessary to assure a sufficiently high resolution of images throughout choosing appropriate pulse sequences. After that, the acquired data can be processed to produce the T2 maps, to portray non-invasive collagen content, architecture of the ROI, changes in the water content (distribution of interstitial water in the solid matrix) and the spatial variation in depth. The aim of this work is firstly to introduce the meaning of T2 relaxation and methods for calculating T2 relaxation times. Further, the aim of this work is to give a brief description of the current pulse sequences used to display menisci.

Keywords: segmentation; T2 relaxation time evaluating; T2 mapping

INTRODUCTION

Meniscus is a structure carrying out several functions in the knee joint. It performs load bearing and so prevents direct contact of the cartilaginous ends of the femur and tibia, reduces stress on the articular cartilage, decomposes evenly pressure on the contact surfaces of cartilages, absorbs shocks and controls external and internal rotation of the tibia. Therefore, it is considered to be a secondary knee stabilizer. Meniscus is a complex hyperviscoelastic biomaterial. Its stiffness is ensured by hydrophilic elements. Their content in meniscus compared to articular cartilage is lower. The permeability of meniscus is affected by the amount, type and arrangement of collagen fibers, mainly represented by the type I. This type of fibers is mechanically much stiffer, able to withstand tension but it has low compressive, flexural and torsional stiffness compared to collagen type II (found eg. in articular cartilage) (Bae, Du, Bydder, & Chung, 2010). That is why meniscus is able to maintain its volume relatively well and shows significant changes of its thickness. Loaded meniscus expands sideways, thus covers still larger area of tibial plateau.

Due to the intra characteristics of the knee joint, the ability to view changes directly in its original position is highly problematic. Magnetic resonance imagining (MRI), an imaging method allowing displaying pathological changes in the human body, is despite the lower signal intensity of the meniscus considered the best non-invasive imaging method (Braun & Gold, 2012). The MR image is formed by the signals from hydrogen protons in water bound to the tissue. The contrast in the MR image depends mainly on the tissue water concentration and its T1 and T2 relaxation times. T1 and T2 relaxation times are influenced by the interaction of the water protons with their surrounding and with each other, respectively, and therefore reflect tissue structure. The values of T1 and T2 times depend on the strength of the magnetic field, e.g. the transition from 1.5 T scanner to 3 T reduce the T2 time to 85% (Stanisz et al., 2005, in Tintera, 2008).

PURPOSE

The aim of this paper is to firstly introduce the basics of MR imaging methods and secondly, to create a brief overview of the current articles dealing with non-invasive imaging of menisci in vivo using MR scanner and using T2 relaxation times to determine the changes in the meniscus.

METHODS

Basic MR imaging measurement sequences and their parameters

Given T1,T2 relaxation times and the concentration of water molecules in the tissue the resulting MR image is highly influenced by type, timing and parameters of the selected measurement sequence (it should be noted that also other tissue parameters such as diffusion or flow influence MR image). Sequence parameters control the image contrast, intensity and

resolution. The most common measurement sequences and dedicated sequences for knee imaging are briefly described below.

Most important parameters of MR imaging sequences

A MR imaging sequence consists of a set of radio-frequency pulses and magnetic field gradients. The selection of the sequence depends on the measured body part and the particular diagnostic question. Sequence parameters control the image contrast, intensity and resolution, therefore, their adjustment plays an important role. Most important parameters of the MR imaging sequence are explained below.

Repetition time (TR) – the time between two consecutive excitation RF pulses. It controls the degree of the image T1 weighting and influences the acquisition time.

Echo time (TE) – the time between the excitation (90°) pulse and the time of the echo occurrence. It controls the degree of T2 weighting. The shorter the time is, the higher the intensity of the echo. However, too short TE causes a loss in the image T2 contrast because tissue T2 curves are too close together. On the contrary, if TE is very long, the echo signal is very weak and the signal to noise decreases.

Field of view (FOV) – represents an imaged area that contains a volume of interest. It is defined as two-or three-dimensional size of the image. With smaller FOV the higher resolution is achieved, however, at the expense of the lower measured signal. The correct choice of the field is important for MR image quality.

Image (acquisition) matrix and resolution – the MRI image is composed of pixels representing signals from tissue volume elements (voxels). The acquisition matrix determines in how many voxels the FOV is divided and, therefore, influences image resolution. Image resolution is of high importance for correct interpretation of MR images. Meniscus, tendons and ligaments have very short T2 times and usually occur in the standard MR image like black areas, which may cause difficulties while evaluating the results. For a reliable in-time diagnosis of lesions ultra structural morphological composition of fibrous tissue high resolution images are essential.

Slice gap (slice spacing) – a distance between two neighboring slices. By reducing the distance between slices it is possible to achieve more precise rendering of the imaged area.

Acquisition time – the time required to gain data from the scanned FOV. This time is restricted due to the patient comfort as well as possible motion artifacts. Acquisition time depends on number of parameters as mentioned before.

Field strength – the strength of the external magnetic field. Scanners can be divided according to the field strength into three groups: low field (0.15 to 1.0 T), medium field (1.0 to 2.0 T) and high field (2.0 to 7.0 T) scanners. The optimal field strength depends on the application**.** As reported by Tintera (2008) “high field systems enable to improve

spatial resolution and fast dynamic examination. On the other hand devices with a weaker field have better tissue contrast. To display cartilage, there has not been demonstrated a clear clinical benefit of field 3 T yet. If the extremely high resolution is required, the time of measurement (without parallel techniques) with no deterioration in quality at 3 T field is about a half compared to 1.5 T for certain sequences."

Open MR devices were developed to monitor tissue in the natural physiological conditions. The devices have magnetic field of lower strength (0.6 T), and so lower image quality and longer acquisition time.

Description of the selected MR imaging sequences

Spin echo sequence, calculation of T2 relaxation time

One of the fundamental pulse sequences used in the magnetic resonance imaging is spin echo sequence (SE) that enables the acquisition of T1, T2 and PD weighted images. The sequence is composed of 90° and the subsequent 180° refocused pulses (Figure 1).

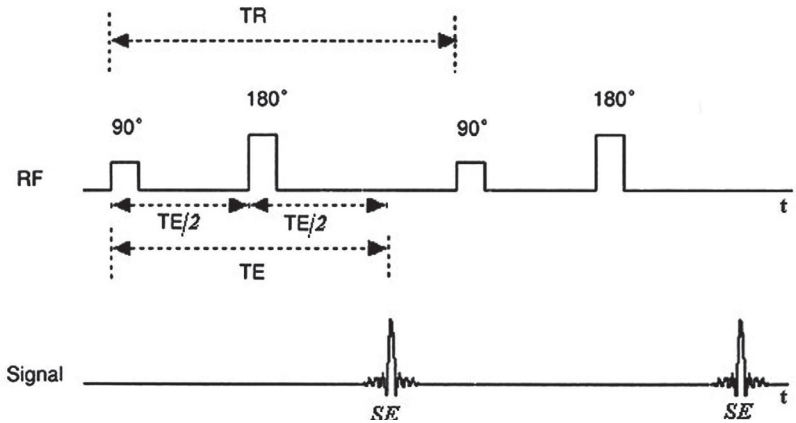

Figure 1. SE sequence consisting of 90° pulse and the subsequent 180° pulse. Figure edited from Fieremans (2009) and subsequently adapted

After the 90° excitation pulse individual protons begin to lose phase coherence by the effect of local inhomogeneities (also called T2* relaxation), resulting in a decrease of transversal magnetization. After some time (TE/2) another, so called refocusing (180°) pulse follows. This pulse causes that the protons start performing the movement with the same precession frequency but in the opposite direction than it was carried before the pulse. As a result, after the period of time equal to TE/2, the protons will be re-phased and restore transversal magnetization. A strong signal which can be detected by the receiver coil at this moment is known as the spin echo (SE). It should be noted that the resulting SE signal is compensated for signal dephasing caused by the external magnetic field inhomogeneity (T2*), however, it is not compensated for T2 relaxation.

In so called multi spin echo sequences several 180° pulses can be employed and multiple echoes recorded. This process can be repeated until the signal decays due to T2 relaxation to zero. The decay of the echo signal S(TE) is exponential as follows
$S(TE) = S(0) * \exp(-TE/T2)$.
Therefore, the acquisition of multiple echoes and the evaluation of measured signals enables the calculation of the T2 time. The calculation is usually performed by the minimization of the square root difference between the measured echo signals and the model function
$S(TE) = A * \exp(-TE/T2) + B$
using Levenberg-Marquardt algorithm.

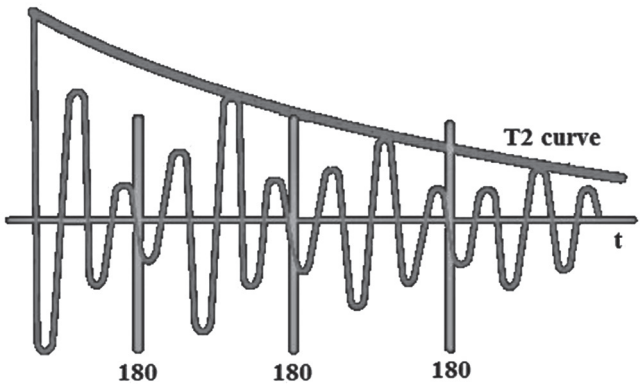

Figure 2. The T2 curve created as a result of the pulse sequence 90°–180°–180°–180°. It shows that the individual echoes are getting gradually lower signal intensity

Turbo spin echo or Fast spin echo (TSE, FSE) – TSE is a modification of the multiple spin echo sequence when the phase encoding gradients are changed for each echo. In this manner the acquisition time can be significantly reduced. This is the reason why these sequences have almost replaced the conventional T2-weighted SE sequences. In T2-weighted TSE images the water and fat appear highly intense.

Gradient echo (GE) – in the gradient echo pulse sequence an additional magnetic field gradient is added to the existing external magnetic field for a very short time. This results in a controlled increase of magnetic field inhomogeneity and a faster decay of the transverse magnetization. After the time TE/2 the polarity of the gradient is reversed. After the next time interval TE/2 an echo, this time called a gradient echo can be observed. GE is similar to the spin echo, however, the echo signal is not T2 weighted but T2* weighted. The longer the TE time is, the more T2* weighted the resulting image becomes.

SPGR (Spoiled Gradient Echo) or FLASH (fast low angle shot) – a gradient echo sequence with a short TR that is mainly used for T1 weighted images (Hashemi, Bradley, & Lisanti, 2012; Mala, 2011). An SPGR sequence spoils the transverse magnetization by semi-randomly changing the phase of the RF pulse (RF spoiling), resulting primarily in

T1 but also PD contrast. For knee imaging a selective radiofrequency (RF) pulse of short-duration (<1 msec) is usually used to excite a slab covering the entire knee joint.

Multiple slice imaging techniques – a concept that can be applied to various MR imaging sequences. The acquisition of a single image may take up to several minutes, depending on the spatial resolution. When searching for pathology, it is usually necessary to obtain several images from different locations. Using conventional approach the acquisition of multiple slices would require multiple repetitions of the whole sequence and in turn very long acquisition time. Multiple slice imaging techniques enable the acquisition of several slices in approximately the same time as required for a single-slice acquisition. During one TR period not only one but several parallel slices are excited and acquired. When TR is longer than TE other slices can be excited and their signal measured while waiting for the recovery of longitudinal magnetization after the signal measurement in the first excited slice. However, neighboring slices acquired during the TR period must be at a sufficient distance from each other so that the excitation of one slice does not affect another slice. It is even possible to measure the whole T2 relaxation curve for the selected slices in a single scan (Herynek, 2013). Although this technique can be used for various pulse sequences it is mostly applied with the SE sequences, because in the SE sequences the TR is much longer than the TE.

There is a limit on the number of slices that can be acquired within a given TR. The limit is given by the time required to excite and collect the echo from each slice: N (max) = TR / (TE + Tc), where N(max) indicates the maximum number of scanned slices and Tc marks the time from the echo peak to the end of the signal (Hendrick, 2008, p. 58).

Ultra short echo time (UTE) sequence – by shortening RF pulses and optimizing the signal acquisition very short TE can be achieved. This is beneficial for imaging structures with very short T2. TE can be up to 100–1000 times shorter in UTE sequences than those used in conventional imaging sequences. As reported by Thakkar, Subhawong, Carrino, and Chhabra (2011) and Qian, Williams, Chu, and Boada (2012), UTE with high resolution allows to display both the surface and the deep layers of connective tissues in the knee. It also displays small structures within the cartilage, meniscus, ligament and patellar tendon. When using high plane resolution (0.28 and 0.14 mm) UTE allows the detection of early damage or fine reparative changes in these tissues. UTE requires a short excitation (0.5 ms) and short acquisition delay (less than 0.2 ms) to reduce TE and thus minimize T2 decay. Since UTE is a relatively new method, research continues to refine 2D and 3D sequences for clinical scanners that would be faster, have better resolution and contrast. For 2D acquisition images with the in plane resolution of 0.3 to 0.8 mm, the slice thickness of 3–5 mm and the total acquisition time 3–17 min have been reported (McWalter, Braun, Keenan, & Gold, 2012). The contrast between the tissue with short T2 relaxation time and surrounding tissues, which may or may not have a short T2 times, can be improved by using several other techniques based on UTE (McWalter et al., 2012).

AWSOS (Acquisition-weighted stack of spirals for fast high-resolution three-dimensional ultra-short echo time MRI) – this sequence based on the spiral data acquisition provides

3D UTE images suitable for knee imaging. It allows the acquisition of 3D images with high resolution (0.28 to 0.14 millimeters) and with the reasonable acquisition time (5–10 min) (Qian et al., 2012).

3D variable TE1 CARTESIAN SPGR – standard Fourier encoded 3D sequences can be adapted to provide the sub millisecond TE using nonselective excitation pulses, highly asymmetric readouts in combination with the variable TE1 along the phase and slice encoding directions. The sequences represent suitable alternative to the morphological musculoskeletal UTE imaging in clinical practice and they have a good potential for biochemical quantification techniques (eg T2* mapping). A minimum TE is typically about 2.8 ms for 0.5 mm in plane resolution in 3D Cartesian scan. The variable TE1 Cartesian methods are fast and flexible and unlike the UTE imaging and provide easy adjustment for the integration of other preparation schemes, such as the suppression of long T2 time components or suppression of the fat signal. (Deligianni, Bär, Scheffler et al., 2012).

T2 relaxation in cartilage and menisci

As stated by Rauscher et al. (2008) in their study, T2 relaxation times reflect more closely the changes in the meniscus compared to the T1 relaxation time, whereas in articular cartilage the opposite is true. Fluids and the tissues with higher water content have long relaxation time T2, which is related to the small size of the water molecules. Their rapid movement causes the time averaging of the local magnetic fields formed by individual water protons. As a result, they do not create significant local magnetic field inhomogeneities that would reduce the T2 relaxation time. In contrast, a mixture of water and organic molecules of greater size (eg. collagen fibers) reduces T2 times. The matrix of the meniscus is composed mainly of highly organized collagen structures in which the most of fibers is oriented in the circumferential direction. Therefore, T2 times in the meniscus are short (Juras et al., 2013; McWalter et al., 2012). T2 times are sensitive to the structure of the ECM, especially on the interaction based on the content, orientation and anisotropy of collagen (Bae et al., 2010; Fragonas et al., 1998; Liess et al., 2002). Damage of the matrix and enhancement of water content in degenerating cartilage can increase T2 relaxation times.

Thanks to the calculation and the display of the T2 maps it is possible to detect the changes in the collagen component of the ECM noninvasive and more precisely to detect the changes in the water content (distribution of interstitial water in the solid matrix) which are not normally visible in conventional MRI images (Nishii et al., 2008; Welsch et al., 2008).

RESULTS

Summary of studies dealing with MR imaging of the meniscus

Described papers were selected using EBSCOhost and Google Scholar services. Also papers found in references of the selected articles were included. The summary of basic information about measurement evaluation methods is provided in the following pages

(Figure 4). Mere comparing the T2 times values stated in individual measurements does not allow making a general conclusion. The reason is that several parameters such as measurement sequence type, final resolution, ROI segmentation technique, the area of interest, age of the subjects, the load history, and the degree of the damage of the meniscus and other differ among individual studies. Moreover, the absolute values of T2 times may depend on the field strength, coil sensitivity, temperature, gradient systems, pulse sequences etc. Therefore the direct comparison of reported T2 relaxation times is problematic. The only correct approach is to compare relative changes in T2 values measured on the same scanner under the same conditions. Generally speaking, T2 relaxation times in healthy meniscus of middle age person range around 11 ms (Rauscher et al., 2008; Stehling et al., 2011; Zarins et al., 2010). Higher values are listed in medial meniscus (Chiang et al., 2013; Subburaj et al., 2012; Tsai et al., 2009; Zarins et al., 2010). Further, T2 times are the lowest in the white zone and they increase in direction to the red zone (Chiang et al., 2013; Tsai et al., 2009).

DISCUSSION

The authors very often use the SPGR pulse sequences to assess the degree of damage. With those sequences they achieve a relatively good resolution (about $0.3 \times 0.3 \times 1$) in a reasonable acquisition time (max. 9 min). 3D SPGR sequences are usually combined with T2 preparation modules. The results show that the lowest acquisition time is around 5 min when using the 3D UTE sequences AWSOS. The segmentation is performed either with a semi-automatic programs written in Matlab or IDL programming languages based on the edge detection or also by the manual detection of the borders. In an effort to reach a low error rate, the measured edges are often completely neglected from the segmentation. T2 relaxation times are mostly calculated using a mono-exponential fitting procedure based on the of the least squares method.

The reason why higher T2 values can been found in the healthy medial meniscus is the different organization of collagen fibers in both parts of the meniscus. In the medial meniscus collagen fibers are more organized, they have higher anisotropy. Also in the full knee extension phase, medial meniscus is loaded by less pressure element in comparison to the lateral meniscus (Athanasiou & Sanchez-Adams, 2009). Because the water is not expelled from the tissue T2 times in medial meniscus are increased.

The reported values indicate also the increase of T2 times depending on the geometric location of the monitored area (red zone – the highest T2 time) and the degree of meniscus damage (the degree of damage increases together with T2). T2 values increase also proportionally to the age. This fact can be explained by increasing free water content and mobility and also by the destruction of the cartilage with increasing age which may in turn lead to changes in meniscus. The studies have also showed that the T2 times in menisci increase after extreme loading. Directly after 30 min running, there are no significant changes in T2 relaxation times in any area of meniscus (Subburaj et al., 2012), however, significant changes were found after 48–72 hours (Stehling et al., 2011). The reason can be that during the resting time, tissue can by swollen by drawing free water into the matrix. Also, released catabolic

substances cause loosening of the collagen network. The permeability of the tissue increases, more interstitial (free) water appears and penetrates into this newly – acquired space.

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REFERENCES

ATHANASIOU, K. A., SANCHEZ-ADAMS, J. (2009). *Engineering the Knee Meniscus* (K. A. Athanasiou Ed. 1st ed.). Rice University. Morgan Claypool Publishers.

BAE, W. C., DU, J., BYDDER, G. M., CHUNG, C. B. (2010). Conventional and ultrashot time-to-echo magnetic resonance imaging of articular cartilage, meniscus, and intervertebral disk. *Top Magn Reson Imaging* 21(5), 275–289.

BRAUN, H. J., GOLD, G. E. (2012). Diagnosis of osteoarthritis: imaging. *Bone* 51(2), 278–288.

DELIGIANNI, X., BÄR, P., SCHEFFLER, K., et al. (2012). High-resolution Fourier-encoded sub-millisecond echo time muskuloskeletal imaging at 3 Tesla and 7 Tesla. *Magnetic Resonance in Medicine* 70(5), 1434–1439.

FIEREMANS, E. (2009). *Validation Methods for Diffusion Weighted Magnetic Resonance in Brain White Matter* (Ph.D. thesis), Gent University, Belgium. Retrieved from: fieremans.diffusion-mri.com/validation-methods-diffusion-weighted-mri-brain-white-matter.

FRAGONAS, E., MLYNÁRIK, V., JELLÚS, V., et al. (1998). Correlation between biochemical composition and magnetic resonance appearance of articular cartilace. *Osteoarthritis and Cartilage* 6(1), 24–32.

HASHEMI, R. H., BRADLEY, W. G., LISANTI, C. J. (2012). *MRI: the basics*. Lippincott Williams & Wilkins.

HENDRICK, R. E. (2008). *Breast MRI: Fundamentals and Technical Aspects*. New York: Springer.

HERYNEK, V. (2013). MR imaging – T1 and T2 relaxation, contrast of MR image, relaxometry. In vivo molecular and cell imaging 2013. Presentation – specialized seminar. ZRIR. IKEM Prague.

CHIANG, S. W., TSAI, P. H., CHANG, Y. C., et al. (2013). T2 Values of Posterior Horns of Knee Menisci in Asymptomatic Subjects. *Plos One* 8(3).

JURAS, V., APPRICH, S., ZBYN, S., et al. (2013). Quantitative MRI analysis of menisci using biexponential T2* fitting with a variable echo time sequence. *Magnetic Resonance in Medicine* 71(3), 1015–1023.

LI, X., MA, B. C., LINK, T., M., et al. (2007). In vivo T(1rho) and T(2) mapping of articular cartilage in osteoarthritis of the knee using 3 T MRI. *Osteoarthritis and Cartilage* 15(7), 789–797.

LIESS, C., LÜSSE, S., KARGER, N., et al. (2002). Detection of changes in cartilage water content using MRI T2-mapping in vivo. *Osteoarthritis and Cartilage* 10(12), 907–913.

MALÁ, A. (2011). Dynamic T1 – contrast enhanced MR imaging. (Master thesis), Masarykova Univerzita, Přírodovědecká fakulta, Brno. Retrieved from: http://is.muni.cz/th/211542/prif_m/dipl_MRI_CA_Mala.pdf.

McWALTER, E. J., BRAUN, H. J., KEENAN, K. E., GOLD, G. E. (2012). Knee. In *Encyclopedia of Magnetic Resonance*.

NISHII, T., KURODA, K., MATSUOKA, Y., et al. (2008). Change in knee cartilage T2 in response to mechanical loading. *J Magn Reson Imaging* 28(1), 175–180.

QIAN, Y., WILLIAMS, A. A., CHU, C. R., BOADA, F. E. (2012). High resolution ultrashot echo time (UTE) imaging on human knee with AWSOS sequence at 3.0 T. *Journal of Magnetic Resonance Imaging* 35(1), 204–210.

RAUSCHER, I., STAHL, R., CHENG, J., et al. (2008). Meniscal Measurements of T1 and T2 at MR Imaging in Healthy Subjects and Patients with Osteoarthritis. *Radiology* 249(2), 591–600.

STEHLING, C., LUKE, A., STAHL, R., et al. (2011). Meniscal T1rho and T2 measures with 3.0T MRI increases directly after running a marathon. *Skeleat Radiol* 40(6), 725–735.

STEHLING, C., SOUZA, R. B., HELLIO LE GRAVERAND, M. P., et al. (2012). Loading of the knee during 3.0T MRI is associated with significantly increased medial meniscus extrusion in mild and moderate osteoarthiritis. *European Journal of Radiology* 81(8), 1839–1845.

SUBBURAJ, K., KUMAR, D., SOUZA, R. B., et al. (2012). The acute effect of running on knee articular cartilage and meniscus magnetic resonance relaxation times in young healthy adults. *The American Journal of Sports Medicine* 40(9), 2134–2141.

THAKKAR, R. S., SUBHAWONG, T., CARRIO, J. A., CHHABRA, A. (2011). Cartilage Magnetic Resonance Imaging Techniques at 3 T: Current Status and Future Directions. *Topics in Magnetic Resonance Imaging* 22(2), 71–81.

TINTERA, J. (2008). MR zobrazování s magnetickým polem 3 T: teoretické aspekty a praktická srovnání s 1,5 T. *Česká Radiologie* 62(3), 233–243.

TSAI, P. H., CHOU, M. C., LEE, H. S., et al. (2009). MR T2 values of the knee menisci in the healthy young population: zonal and sex differences. *Osteoarthritis and Cartilage* 17(8), 988–994.

WELSCH, G. H., TRATTNIG, S., SCHEFFLER, K., et al. (2008). Magnetization transfer contrast and T2 mapping in the evaluation of cartilage repair tissue with 3T MRI. *Journal of Magnetic Resonance Imaging* 28(4), 979–986.

WILLIAMS, A., QIAN, Y., GOLLA, S., CHU, C. R. (2012). UTE-T2* mapping detects sub-clinical meniscus injury after anterior cruciate ligament tear. *Osteoarthritis and Cartilage* 20(6), 486–494.

ZARINS, Z. A., BOLBOS, R. I., PIALAT, J. B., et al. (2010). Cartilage and meniscus assesment using T1rho and T2 measurements in healthy subjects and patients with osteoarthritis. *Osteoarthritis and Cartilage* 18(11), 1408–1416.

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Table 1. The authors who have worked on the visualization of the knee meniscus for last 6 years: a summary of used sequences and procedures, basic imaging parameters and segmentation techniques, method for calculation of T2 relaxation times and their resulting values (sign "?" means that value was not stated in published paper

Author (Year) Object of work	Used sequences and parameters		Number of subjects (N) and Segmentation (S)	Calculation of T2 relaxation times	Making T2 maps, the resulting values of T2 relaxation times ± SD
Rauscher et al. (2008) Sequences according to Li et al. (2007) comparison of T2 times meniscus healthy and those with OA	*SPGR sequences*		N: 23 healthy subjects (age 34.1 ± 10), 27 with mild OA (age 52.5 ± 10.9), and 10 with severe OA (age 61.6 ± 11.6). S: Program in Matlab and use of semi-automated technique based on the edge detection and parametric Bezier curves.	The meniscal segmentations were then resampled and superimposed onto the T1rho and T2 maps to define the ROI for T1rho and T2 assessment.	T2 maps were automatically coregistered to the SPGR images.The selective T2 preparation sequences were added into the SPGR sequences and 3D sagittal T2 maps were created. After segmentation, the meniscus was transformed into a 3D binary mask with isotropic voxels, and T2 maps were reconstructed Values for both menisci: Health 11.4 ms ± 3.9 ms, middle level OA 13.5 ms ± 4.7 ms and a severe OA 16.6 ms ± 8.2 ms.
	TR/TE (ms)	20/7.5			
	FOV (mm)	160 × 160			
	Matrix	512 × 256			
	voxel size	0.293 × 0.293 × 1			
	aq. time (min)	7:37			
	non-selective T2 preparation sequences				
	TR/TE (ms)	2000/4.1; 14.5			
	FOV (mm)	140 × 140			
	Matrix	512 × 512			
	voxel size	0.54 × 0.72 × 3			
	aq. time (min)	10:36			
Tsai et al. (2009) zonal and also sex differences in the back corner of med. and lat. meniscus in 3 zones	*multislice TSE sequences*		N: 10 men and 10 women (age 26.5 ± 2.74). S: ROIs identified on the sagittal MR images of the first echo in the posterior horns of the med. and lat. menisci. To avoid partial volume effects, the upper and lower borders of the meniscus were not selected in the ROIs.	T2 were derived using the least square single-exponential curve-fitting method in the Matlab. Analysis of the T2 values was conducted on a zone-by-zone basis. The mean signal intensities were calculated in the ROIs of the posterior horns of the med. and lat. menisci on each slice of the motion-corrected images.	T2 maps are not produced. Zonal differences found in the back corner of the meniscus: inner white zone T2 = 8.02 ms ± 0.60 ms, white and red T2 = 8.78 ms ± 0.99 ms, outer red zone T2 = 22.12 ms ± 0.92 ms.
	TR/TE (ms)	2500/6.4; 9.4; 12; 15			
	FOV (mm)	?			
	Matrix	258 × 324			
	voxel size	? × ? × 1			
	aq. time (min)	14:40			

Zarins et al. (2010) value of T2 relaxation times in determined part of the meniscus depended on the damage	*3D SPGR*		N: 29 men, 34 women (age 51 ± 13.6). 19 of them without damaging the meniscus 26 with moderate OA and 18 with a severe OA. S: according to Raucher (2008) on the SPGR high-resolution images was performed semi-automatic segmentation at home created software in Matlab based on the edge detection and Bezier parametric curves.	All images were transferred to a Sun Workstation to create the T2 maps. The last T2-weighted image had extremely low signal, therefore, was not used for the reconstruction of T2 map.	Reconstructed by fitting the T1rho and T2 images pixel by pixel using a at home-made LM algorithm. A significant increase of the time in the posterior corners of med. and lat. meniscus at a higher level of damage was founded. Lat. meniscus: a healthy 10.42 ms ± 1.45 ms, higher levels of damage 12.03 ms ± 2.12 ms med meniscus: a healthy 11.26 ms ± 1.51 ms, corrupted 17.89 ms ± 4.51 ms.
	TR/TE (ms)	15/6.7			
	FOV (mm)	140 × 140			
	Matrix	512 × 512			
	voxel size	0.273 × 0.273 × 1			
	aq. time (min)	7:37			
	T2 preparation sequences				
	TR/TE (ms)	9. 3/3.1; 13.5; 23.9; 44.8			
	FOV (mm)	140 × 140			
	Matrix	256 × 192			
	voxel size	0.54 × 0.73 × 3			
	aq. time (min)	11:00			
Stehling et al. (2011) meniscus measured 48–72 h after completing marathon	*SPGR sequences*		N: The experimental group – 8 women and 5 men (age 32.3 ± 5.6), the control group – 4 men and 6 women (age 30.5 ± 5.3 years). S: in-house software developed with an Interactive Display Language (IDL). An IDL routine was used to simplify the drawing of splines delineating meniscal areas.	After the segmentation IDL routine was used to calculate the average T2 values of on the ROI generated maps.	Images were transferred to a remote SUN/SPARC station to create T2 maps. Resulting values: significant changes only in runners before competition 11.15 ms ± 1.46 ms, within 3 days after running 13.36 ms ± 1.27 ms, $p < 0.0001$ and 3 months after the competition 11.47 ms ± 1.42 ms.
	TR/TE (ms)	20/7.5			
	FOV (mm)	160 × 160			
	Matrix	512 × 512			
	voxel size	0.293 × 0.293 × 1			
	aq. time (min)	7:37			
	non-selective T2 preparation sequences				
	TR/TE (ms)	2000/4.1; 14.5; 25; 45.9			
	FOV (mm)	140 × 140			
	Matrix	256 × 192			
	voxel size	0.55 × 0.73 × 3			
	aq. time (min)	10:36			

Qian et al. (2012) Meniscus and its display with the highest possible resolution in a short time till 10 min	*Home developed 3D UTE seq. AWSOS*		N: 7 men and 2 women (age 28.3 ± 5.5), 5 of them asymptomatic and 4 with ACL injury. S: demonstration of the image acquisition with high resolution (0.28 mm) in a relatively short time (5–10 min). Meniscus non-segmented.	T2 times were not calculated.	T2 maps were not created.
	TR/TE (ms)	80/0.6			
	FOV (mm)	140 × 140			
	Matrix	512			
	voxel size	0.28 × 0.28 × 2			
	aq. time (min)	5:12 Spiral readout 16.80 ms			
Subburaj et al. (2012) articular cartilage + meniscus front and back corner after 30 min run	*T2 FSE*		N: 10 men and 10 women (age 22–35). S: from the sagittal SPGR images by using in house-made software based on spline-based semi-automatic segmentation algorithm in Matlab (automatic edge detection and manual correction).	Images were transferred to a workstation HP for the off-line quantification of T2 relaxation times.	T1rho and T2 maps were obtained by a combination of T1rho/T2 quantification sequences during post processing. Home made LM algorithm was used to fit the intensity images into the equation. T2-weighted images with TE = 54.8 ms had a low SNR ratio (SNR; <5), therefore they were not included into the reconstruction. Generally higher T2 values were found in medial anterior horn and medial body. After run, T2 times were increased in all regions except the posterior horn of the medial meniscus, but changes were not significant.
	TR/TE (ms)	4300/51			
	FOV (mm)	140 × 140			
	Matrix	512 × 256			
	voxel size	0.27 × 0.54 × 0.5			
	aq. time (min)	?			
	sag 3D SPGR; combined T1rho/T2 quantification sequences				
	TR/TE (ms)	15/6.7; 0/13.67/27.34/54.68			
	FOV (mm)	140 × 140			
	Matrix	512 × 512; 256 × 128			
	voxel size	0.27 × 0.27 × ?; 0.54 × 1.09 × 3			
	aq. time (min)	9:00			

Stehling et al. (2012) meniscal extrusion by applying loading 50% of body weight	*Sag.+coronal 2D T2-weighted FSE*		N: 10 healthy subjects and 20 with confirmed OA (age less than 40 years).	Qualitative change was not followed, only a shift of the burden meniscus.	Degenerative knee abnormalities demonstrate an increase of meniscal extrusion in a loaded knee joint compared to healthy samples.
	TR/TE (ms)	4000/48; 3000/10			
	FOV (mm)	140 × 140			
	Matrix	384 × 192			
	voxel size	0.36 × 0.72 × 2			
	aq. time (min)	?			
	coronal 3D SPGR				
	TR/TE (ms)	22/7.0			
	FOV (mm)	140 × 140			
	Matrix	512 × 512			
	voxel size	0.27 × 0.27 × 1.5			
	aq. time (min)	?			
Williams, Qian, Golla, and Chu (2012) T2* values of menisci in healthy subjects compared to subjects ACL rupture or both ruptured ACL and meniscus tearing	*AWSOS*		N: 10 healthy subjects, 25 with a ruptured ACL (age 26–77 years). S: manual rendering of the posterior horn of med. and lat. meniscus on the AWSOS sequences obtained at TE = 7 ms, where the contrast between the meniscus and the surrounding tissue was strongest.	Interpolation: Prior to T2 curve-fitting, TE images from the AWSOS sequence collected in vivo were linearly interpolated to a matrix size of 512 (or a pixel size of 273 mm) to permit finer image registration.	UTE-T2* maps were generated with a mono-exponential T2 curve-fit of all 11 echo images using MRI Mapper software. Values T2: in asymptomatic individuals 8 ms ± 2 ms, patients without rupture of the meniscus but ruptured ACL 13 ms ± 5 ms and patients with ruptured meniscus and the ACL 21 ms ± 7 ms.
	TR/TE (ms)	0.6; 1; 2; 3; 4; 5; 7; 10; 20; 30; 40			
	FOV (mm)	140 × 140			
	Matrix	256 × 256 lin. interpolated to 512 × 512			
	voxel size	0.273 × 0.273 × ?			
	aq. time (min)	22:00			

Chiang et al. (2013) Age, sex and zonal differences in the posterior corners of the meniscus	*multislice TSE sequences*		N: 30 asymptomatic men and 30 asymptomatic women (age groups 20–34, 35–49, 50–70 years). S: meniscus was divided into red, white and white-red according to the vascularization. On the omission of partial volume effects, the upper and lower edges were not included in the ROI.	T2 were derived from each image and averaged over all slices in on the ROI on the motion-corrected image. Then T2 values were derived using the least squares method in Matlab. The accuracy of fitting is reported by the R^2 values used in the non-linear curve fitting.	Motion correction: 2D auto-correlation method was used, in which the first image acquired with TE = 6.4 ms was used as a reference image to coregister the other three echo images acquired with TE = 9.4, 12, 15 ms, and between-slice 2D correlation coefficients were calculated. Results: 9.94 ms ± 0.94 ms, 10.73 ms ± 1.55 ms and 12.36 ms ± 2.27 ms for women of different ages. For men: 9.17 ms ± 0.74 ms, 9.64 ms ± 0.67 ms and 10.95 ms ± 1.33 ms.
	TR/TE (ms)	2500/6.4; 9.4; 12; 15			
	FOV (mm)	258 × 324			
	Matrix	?			
	voxel size	?			
	aq. time (min)	14:40			
Juras et al. (2013) T2* times of menisci	*3D vTE Cartesian SPGR sequences*		N: 48 healthy menisci, 12 degenerated and 8 with tears (study includes together 8 men age 34 ± 10 and 9 women aged 36 ± 14). S: ROI were defined on the five successive sections, each section was divided into two regions (white and black).	T2*: Images from the vTE sequence were analyzed using a custom-written script in IDL. A monoexponential as well as a biexponential fitting procedure was employed on all MR data sets on a pixel-by-pixel basis.	Comparison of the ability of monoexponential (Me) and biexponential (Be) T2* fitting to distinguish between healthy, degenerated and torn meniscus. Me: healthy 7.61 ms ± 2.49 ms, degenerated 9.54 ms ± 2.25 ms, meniscal tears 14.59 ms ± 5.24 ms. Be (compare short/long components of T2*): healthy 0.82 ms ± 0.38 ms / 15 ms ± 5.4 ms, degenerated 1.29 ms ± 0.53 ms / 19.97 ms ± 5.59 ms, meniscal tears 2.05 ms 0.73 ms / 26.83 ms ± 7.72 ms.
	TR/TE (ms)	29/0.75; 3.51; 5.87; 8.23; 10.6; 12.96; 15.33; 17.69; 20.06; 22.42			
	FOV (mm)	120 × 180			
	Matrix	256 × 176			
	voxel size	0.47 × 1.02 × 0.7			
	aq. time (min)	12:16			