The British Journal of Radiology, 71 (1998), 31–36

The magic angle phenomenon in tendons: effect of varying the MR echo time

W C G PEH, FRCR and J H M CHAN, MSc

Department of Diagnostic Radiology, The University of Hong Kong, Queen Mary Hospital, Hong Kong

Abstract. Increased signal intensity on magnetic resonance (MR) imaging of tendons arising from the magic angle phenomenon is well recognized. This study aimed to evaluate the effect of varying the echo time (TE) upon tendon signal intensity, and to determine if a modified TE value produces acceptable T_1 and proton density (PD) weighted images. Fresh bovine tendons were imaged in a 1.5 T MR scanner using spin echo (SE) T_1 and PD weighted sequences and utilizing a number of different coils. For each set of sequences, the tendon was orientated at 55° to the main magnetic field (B_0) and imaged using constant TR and incremental TE values. Signal intensity was measured on images at each TR/TE value and compared with the signal intensities of tendons orientated at 0° to B_0, obtained using minimum TE values. This experiment was repeated with a 1.0 T MR scanner and utilizing a spine coil. The Achilles tendon of a human volunteer was similarly imaged using a general purpose flex coil. For bovine and human tendons orientated at 55° to B_0, the signal intensities decreased exponentially with increasing TE. A critical TE value exceeding 37 ms, for each sequence, reduced the signal intensities to the levels obtained with the tendons orientated at 0° to B_0, such that the magic angle phenomenon could be avoided. Although there was variability of the signal intensities with different coils, the critical TE value remained constant and the anatomical clarity was not degraded. The critical TE value was unaltered using two MR scanners of different field strengths.

In MRI, the magic angle phenomenon causes increased signal on short echo time (TE) images. It affects tissues with well-ordered collagen fibres in one direction, such as tendon or articular hyaline cartilage, which both behave in an anisotropic manner in a magnetic field. The induced hyperintense signal is dependent on the orientation of the tissue to the main magnetic field (B_0), with maximum signal intensity observed at the "magic angle" of 54.74° (usually approximated to 55°) to B_0 [1–3].

In clinical practice, it is important to distinguish effects of the magic angle phenomenon in normal tissue from signal abnormalities due to degeneration or partial tear. Suggested methods include close comparison with T_2 weighted images, looking for other features of tendonopathy and repositioning the patient to avoid the magic angle phenomenon [4, 5]. Most of the previous studies of this phenomenon have evaluated the influence of either tendon or articular cartilage orientation upon their MR appearances [6–8].

The purpose of this investigation was to study the effect of systematically varying the MR TE values upon the increased signal intensity observed with the magic angle phenomenon in tendons. We hypothesized that there was a "critical" TE value on both T_1 and proton density (PD) weighted images, above which the magic angle phenomenon should not be apparent. It was hoped that clinical application of modified TE values might reduce the number of false-positive diagnoses of tendon disease.

Received 25 July 1997 and in revised form 9 September 1997, accepted 20 September 1997.

Materials and methods

In vitro experiments were conducted using bovine Achilles tendons. The tendons were fresh and intact, with the distal portions of the gastrocnemius and other contributing calf muscles still attached. A total of six tendons were used, each for less than 2 h at room temperature to ensure freshness. The same tendon was used in each series of experiments in order to achieve consistency in comparison. MR imaging was performed using a 1.5 T clinical whole-body scanner with the 5.4 version software (Signa Advantage, General Electric Medical Systems, Milwaukee, WI, USA). The same settings of centre frequency, transmitter gain and receiver gain were used for each set of sequences.

In the first *in vitro* experiment, the tendon was placed on a plastic sheet and covered by ultrasonic transmission gel (Aquasonic, Parker Laboratories Inc., Orange, NJ, USA). The purpose of the ultrasonic gel was to provide contrast with the anticipated low signal of the tendon and assist the

tuning procedure of the MR system. Imaging was performed using a 5-inch general purpose (GP) circular coil, placed beneath the plastic sheet (Figure 1). Parameters used were 20×20 cm field-of-view (FOV), 256×192 matrix, one excitation, 16 kHz bandwidth and 4 mm interleaved section thicknesses.

The tendon was placed in the centre of the gantry, with its long axis orientated at 0° to B_0. Following a spin echo (SE) T_1 weighted (300/13 [TR/TE]) coronal localizer image, true axial SE T_1 weighted (500/13 [TR/TE]) images were obtained. The same tendon was then re-orientated so that its long axis was 55° to B_0. The angle was checked initially using a protractor and then by measurement on the MR console monitor after obtaining the coronal localizer image. Oblique sections were performed to obtain a series of nine true axial SE T_1 weighted images of the tendon. The TR of these images were kept constant at 500 ms, while the TE was varied by 3 ms intervals for each series from 13 ms to 37 ms.

Quantitative evaluation of tendon signal was measured by manually placing a region of interest (ROI) cursor in the centre of a selected axial tendon image. Care was taken to ensure that the ROI was of the same size and shape, and that the same site was measured on anatomically corresponding axial slices. Measurements were made with the tendon orientated to 0° to B_0 (500/13 [TR/TE]), and at 55° to B_0 (500/13–500/37 [TR/TE]). The *in vitro* experiments using T_1 weighting were repeated using GP flex coils, phase-array spine coils, head, pelvis and body coils. The experiment was also performed, using GP 5-inch and phase-array spine coils, to obtain conventional SE PD-weighted images with TE values varying by 4 ms intervals from 2000/12 to 2000/52 [TR/TE].

The *in vitro* experiment was repeated using a 1.0 T clinical whole-body scanner with the 5.5 version software (Signa Horizon, General Electric Medical Systems, Milwaukee, WI, USA). A spine coil was used to image the bovine tendon. SE T_1 weighted images were obtained with TE values varying by 3 ms intervals from 13 ms to 40 ms, and SE PD weighted images with TE values varying by 4 ms intervals from 12 ms to 40 ms.

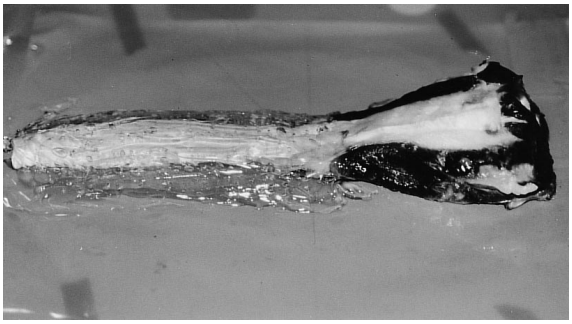


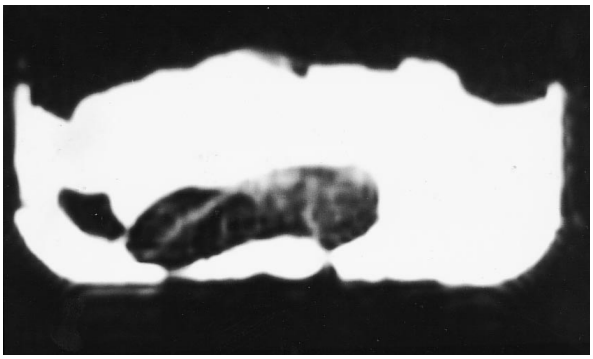
Figure 1. Bovine tendon covered with ultrasonic transmission gel. The 5-inch general purpose circular coil is just visible beneath the plastic sheet.

The *in vivo* experiment was carried out on the Achilles tendon of a healthy 33-year-old human volunteer with no history of ankle or foot problems. Imaging was performed using a GP flex coil with the following parameters for SE T_1 weighted images: 20×20 cm FOV, 256×192 matrix, one excitation, 16 kHz bandwidth and 3 mm interleaved section thicknesses. These parameters were the same for SE PD weighted images, except that a 256×160 matrix was used instead. SE T_1 weighted (500/13 [TR/TE]) and SE PD weighted (2000/12 [TE/TR]) true axial images were obtained, following a SE T_1 weighted (300/12 [TR/TE]) sagittal localizer, with the long axis of the Achilles tendon orientated at 0° to B_0. Using the same method as the *in vitro* experiment, the patient's foot was then re-orientated at 55° to B_0. Ten true axial SE T_1 weighted images were acquired at 3 ms intervals from 500/13 to 500/40 [TR/TE], while seven true axial SE PD weighted images were acquired at a constant TR value of 2000 ms and the following TE values: 12, 20, 28, 32, 36, 40 and 48 ms.

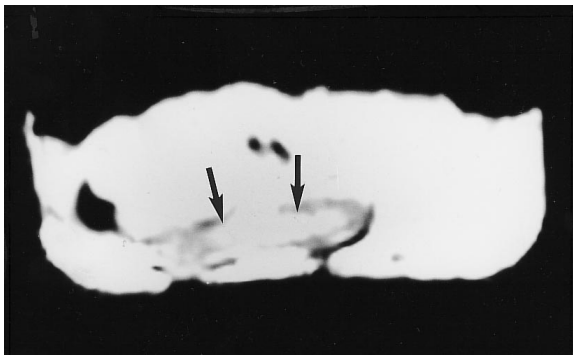
Objective evaluation of tendon signal was made, using the methods described for the *in vitro* experiments. In addition, for each selected T_1 and PD weighted image obtained at 55° orientation, muscle, bone marrow and subcutaneous fat were subjectively assessed for anatomical clarity on the MR console monitor. These were compared with corresponding T_1 and PD weighted images, obtained with the tendon at 0° orientation. Using ROI cursors of the same size and shape, quantitative measurements of signal intensity of bone marrow, subcutaneous fat, muscle and Achilles tendon were made, at the same sites on corresponding T_1 weighted images (500/13 to 500/40 [TR/TE]).

Results

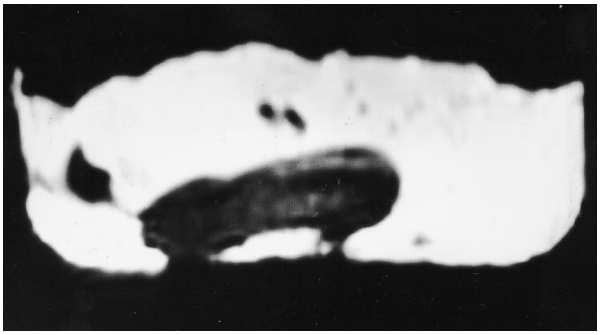
As expected, the signal intensities of both bovine and human Achilles tendons remained low when using minimum TE values on both SE T_1 and PD weighted images with tendon orientation at 0° to B_0. When the tendons were re-orientated at 55° to B_0, the same T_1 and PD weighted TR/TE values produced markedly increased intratendon signal intensity, due to the magic angle phenomenon (Figures 2a and b). Signal intensity decreased in an exponential manner with increasing TE values on both T_1 and PD weighted images of the bovine tendons (Figure 3). If the tendon signal intensity measured at 0° orientation was taken as being normal, then a "critical" TE value of about 37 ms or more was required to reduce the magic angle



(a)



(b)



(c)

Figure 2. Axial SE T_1 weighted images of bovine tendon. The same window level and width was used for all images. (a) Tendon at 0° to B_0 (500/13) shows normal tendon signal. (b) Tendon at 55° to B_0 (500/13) shows increased signal (arrowed) due to the magic angle phenomenon. (c) The high intratendon signal is markedly reduced with alteration of the TE time (500/37) at the same 55° to B_0 orientation.

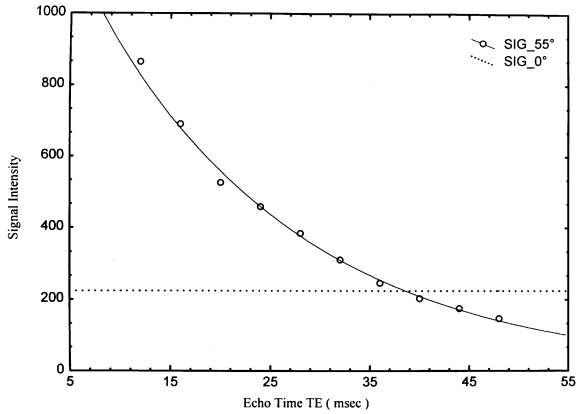


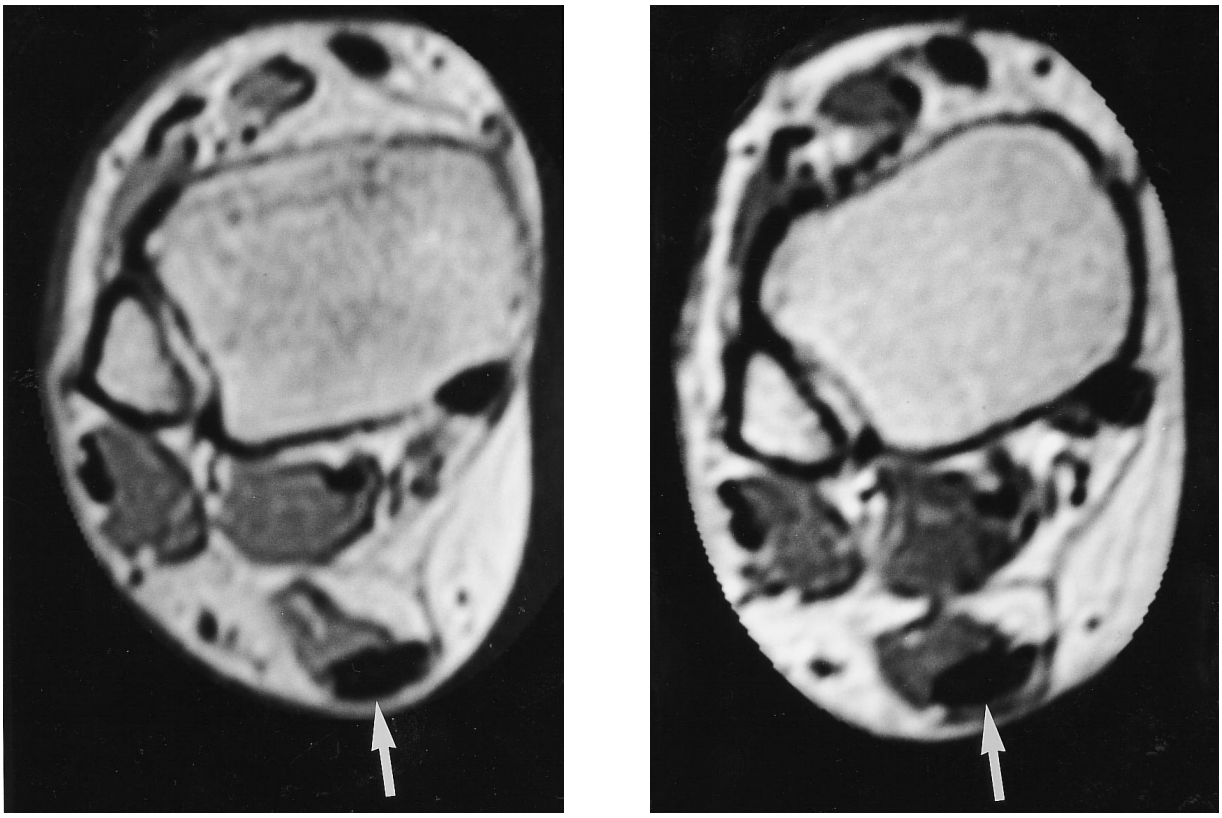
Figure 3. Relative bovine tendon signal intensities plotted against different SE PD weighted TE values, with tendon orientated at 55° to B_0 (bold curved line). TR is constant at 2000 ms. The signal intensity obtained with the same tendon orientated at 0° to B_0, with a SE PD weighted (2000/12) sequence, is shown by the dotted straight line. The two lines intersect at a TE of about 37 ms.

Table 1. Effect of different coils on bovine tendon signal intensity (TR/TE = 500/13 ms)

Coil	Signal intensity at 0° to B_0	Signal intensity at 55° to B_0	Ratio of signal increase at 55° to B_0
Body	53	141	2.7
Spine	70	243	3.5
5 inch GP	72	285	3.9
Head	58	219	3.8
GP Flex	72	288	4.0
Pelvic	70	274	3.9

induced signal intensity near to that of a tendon orientated at 0° to B_0 (Figure 2c). Similarly-shaped exponential decay curves were achieved with the various other coils, with a lower (1.0 T) field strength scanner and with the human *in vivo* experiment. Although the different coils produced different signal intensities and ratios of signal increase, varying from 2.7 for the body coil to 4.0 for the GP flex coil (Table 1), the "critical" TE value of 37 ms remained statistically constant with a standard deviation of 1.2 ms for all coils.

With the appropriate windowing, SE T_1 and PD weighted images of the human tendon orientated at 55° to B_0, obtained at TE values just exceeding the critical level of 37 ms, provided similar anatomical information to SE T_1 and PD weighted images of the tendon orientated at 0° to B_0 acquired with minimum TE values of 13 ms and 12 ms, respectively. With adjustment of the window width and levels to optimize anatomical clarity, the subcutaneous fat, bone marrow and muscle had similar appearances on corresponding SE T_1 and PD weighted images, despite differences in TE values (Figure 4). At the critical TE of 37 ms, signal intensities of the SE T_1-weighted images of the human tendon showed 41% reduction for fat (bone marrow and subcutaneous fat), 56% reduction for muscle but an 81% reduction for tendon (Table 2). In both T_1 and PD weighted images of tendon, signal-to-noise ratio (SNR) decreased by about 40% at the critical TE values (Table 3).



(a) (b)

Figure 4. Axial MR images of the human volunteer, showing similar anatomical clarity with short and long TE. (a) SE PD weighted short TE (2000/12) image with tendon orientated at 0° to B_0. (b) SE PD weighted long TE (2000/40) image with tendon orientated at 55° to B_0. The Achilles tendon is arrowed.

Table 2. Comparison of signal intensities of subcutaneous fat, bone marrow, muscle and Achilles tendon with various echo times (TE) on SE T_1 weighted (constant TR of 500 ms) axial images of a normal human ankle

TE (ms)	Signal intensity			
	Subcutaneous fat	Bone marrow	Muscle	Tendon
13	610	560	159	114
16	568	527	154	82
19	537	504	133	68
22	513	471	125	54
25	482	436	108	45
28	453	406	96	34
31	404	373	85	31
34	374	346	73	24
37	357	331	70	22
40	337	310	67	16

Table 3. Calculated decrease in signal-to-noise ratio (SNR) with increasing TE values on T_1 and PD weighted images of a normal human Achilles tendon

TE	SNR	% DROP
(a) T_1 weighted images (TR = 500 ms)		
13	57	0%
16	55	4%
19	52	9%
22	48	16%
25	41	28%
28	39	32%
31	36	37%
34	35	39%
37	**32**	**43%**
40	29	49%
(b) PD weighted images (TR = 2000 ms)		
12	87	0%
20	70	19%
28	61	30%
32	57	35%
36	**51**	**41%**
40	48	45%
48	43	51%

Discussion

Normal tendons produce little or no signal on conventional MRI sequences. This is because the tendon consists of type I collagen fibres orientated in a parallel manner into highly ordered bundles. Tendons, therefore, display structural anisotropy, creating a static local magnetic field which contributes further to spin–spin interaction, causing normal tendons to have ultra-short T_2 relaxation times in the order of 250 μs. The resultant rate of T_2 decay is so rapid that a very low signal is obtained irrespective of TE value [1–3].

The static local magnetic field component B_z is superimposed onto the static field B_0 of the magnet,

causing the spins to precess at different Larmor frequencies, and resulting in spin dephasing. The rate of spin dephasing caused by the anisotropic structure of tendons is thus proportional to a factor $(3\cos^2\phi - 1)$, where ϕ is the angle between the static field of the magnet and the longitudinal axis of the tendon. This additional spin dephasing vanishes if $3\cos^2\phi - 1 = 0$, where $\phi = 54.74^\circ$. In other words, when $\phi = 54.74^\circ$, the spin–spin interaction due to the static local field becomes zero and the T_2 decay is governed by the time-varying field only. Consequently, the T_2 relaxation time rises to about 20 ms and the rate of T_2 decay is less rapid, so that the normal tendon appears hyperintense at short TE values [1–3].

On the other hand, the T_1 relaxation time is independent of tendon orientation. The signal intensity is governed by the same T_2 relaxation time decay curve regardless of whether there is T_1, PD or T_2 weighting. For each type of sequence, the TR (which is constant) and T_1 relaxation time (which is constant and independent of tendon orientation) values would only determine the initial signal intensity, hence the pattern of T_2 decay is unaffected. Thus, regardless of whether the sequences are T_1 or PD weighted, the point in the graph where the T_2 decay curve of a tendon at 55° orientation intersects the horizontal line of a tendon at 0° orientation remains unchanged (Figure 3).

The clinical significance of the "magic angle" ($\phi = 55^\circ$) phenomenon lies in its ability to produce increased signal intensity in normal tendons. The overlap among MRI of the normal rotator cuff tendon, tendinosis and partial tear makes differentiation difficult in some cases [9]. Increased signal on short TE images has been attributed to early tendinosis or tendon degeneration [10]. This pattern of signal alteration is supported by Kjellin et al who showed that MR correlated well with the histological findings of tendon degeneration in cadavers [11].

The magic angle signal artefacts have been recognized at several sites, particularly in tendons of the wrist, ankle and rotator cuff, as well as cartilaginous structures such as the knee meniscus and glenoid labrum. Besides evaluating secondary signs of tendonopathy such as tendon thickening, shape changes and presence of tendon sheath fluid, most authors have advocated repositioning the tendon in question in order to eliminate the magic angle effect [2, 4, 5, 7, 8]. We have shown that the tendon signal changes produced by the magic angle phenomenon can be greatly reduced by increasing the echo time to above a certain critical value. A critical TE value of about 37 ms was obtained in the experiments with both fresh bovine and normal human volunteer tendons, using T_1 as well as PD weighted sequences. The critical TE value remained constant while using different coils and with MR scanners of different field strengths. This critical TE value was independent of the differences in tendon signal intensity produced among the coils used. It was considered unnecessary to scan a large number of animal or human tendons as our hypothesis of a critical TE value could be consistently demonstrated on the generated exponential decay curves plotting signal intensity against TE in each series of experiments.

Imaging the ankle of a normal human volunteer showed that anatomical information was not lost with increasing TE up to the critical TE value. If increased signal from the magic angle phenomenon is suspected on routine short TE sequences, for example where a tendon alters direction across a joint, repeating the T_1 or PD weighted sequence with an increased TE value (of 37 ms or more) should be helpful in distinguishing increased signal due to the magic angle phenomenon from degeneration, tendinitis or partial tear. These and other abnormalities of tendon would be expected to remain visible, if not more prominent, with increased and long TE values. The decrease in SNR with increased TE values can be compensated by doubling the number of excitations in the repeat scan. If the magic angle phenomenon is present, the need to reposition the patient or structure of interest can also be avoided using this method. Alternatively, one may possibly prospectively perform T_1 and PD weighted images with TE values just exceeding 37 ms instead of minimum TE values, if the magic angle phenomenon is anticipated from the position of certain tendons and cartilaginous structures. It would be useful to conduct prospective studies comparing the appearances of various types of tendon lesions in the clinical setting, employing both standard TE and increased "critical" TE values.

Acknowledgment

We thank Ms Alice Lau for her secretarial assistance.

References

1. Fullerton GD, Cameron IL, Ord VA. Orientation of tendons in the magnetic field and its effect on T_2 relaxation times. Radiology 1985;155:433–5.
2. Erickson SJ, Cox IH, Hyde JS, Carrera GF, Strandt JA, Estkowski LD. Effect of tendon orientation on MR imaging signal intensity: a manifestation of the "magic angle" phenomenon. Radiology 1991;181:389–92.
3. Erickson SJ, Prost RW, Timins ME. The "magic angle" effect: background physics and clinical relevance. Radiology 1993;188:23–5.
4. Peterfy CG, Linares R, Steinbach LS. Recent advances in magnetic resonance imaging of the musculoskeletal system. Radiol Clin North Am 1994;32:291–311.

5. Hayes CW, Parellada JA. The magic angle effect in musculoskeletal MR imaging. Topics MRI 1996;8:51–6.
6. Rubenstein JD, Kim JK, Morava-Protzner I, Stanchev PL, Henkelman RM. Effects of collagen orientation on MR imaging characteristics of bovine articular cartilage. Radiology 1993;188:219–26.
7. Peterfy CG, Janzen DL, Tirman PFJ, van Dijke CF, Pollack M, Genant HK. "Magic-angle" phenomenon: a cause of increased signal in the normal lateral meniscus on short-TE MR images of the knee. AJR 1994;163:149–54.
8. Timins ME, Erickson SJ, Estkowski LD, Carrera GF, Komorowski RA. Increased signal in the normal supraspinatus tendon on MR imaging: diagnostic pitfall caused by the magic-angle effect. AJR 1995;164:109–14.
9. Robertson PL, Schweitzer ME, Mitchell DG, Schlesinger F, Epstein RE, Frieman BG, et al. Rotator cuff disorders: interobserver and intraobserver variation in diagnosis with MR imaging. Radiology 1995;194:831–5.
10. Weatherall PT, Crues JV III. Musculotendinous injury. MRI Clin North Am 1995;3:753–72.
11. Kjellin I, Ho CP, Cervilla V, Haghighi P, Kerr R, Vangness CT, et al. Alterations in the supraspinatus tendon at MR imaging: correlation with histopathologic findings in cadavers. Radiology 1991;181:837–41.