Effects of capillary orientation on muscle T2/T2*: comparison of numerical simulations with empirical data

O. A. Sanchez[1], A. M. Heemskerk[1], and B. M. Damon[1]

[1]Radiology and Radiological Sciences, Vanderbilt University Institute of Imaging Science, Vanderbilt University, Nashville, TN, United States

Introduction

Changes in the muscle capillary orientation (α) with respect to B_0 may affect the signal intensity in muscle functional MRI via extravascular BOLD effects (1, 2). Since capillaries run parallel to the longitudinal length of the muscle fibers, the fiber orientation also reflects α. In the soleus muscle, a maximum voluntary contraction modifies the pennation angle of the soleus muscle from 21° to 40° (3). Therefore, contraction-induced signal intensity changes might be explained not only by modifications in physiological or metabolic variables within a muscle, but also by changes in α. Therefore, the purposes of this study were to determine the independent effects of α on T_2 and T_2^* and to compare empirical findings with numerical simulations.

Methods

Protocol: Water diffusion properties, T_2, and T_2* were measured in the soleus muscles of four healthy subjects (2 male). The subjects were studied at 3 different knee angles of 0°, 5° and 10°, where 0°= full extension. In each position, the ankle was maintained at an angle of 90°.

MRI data acquisition and analysis: MRI data were obtained on a 3T Phillips Intera Achieva MR Imager/Spectrometer. A pair of 15x10 cm (length x width) surface coils placed over the medial and lateral heads of the gastrocnemius muscle. Following the acquisition of the 3-plane scout images, the sagittal slices were used to identify the maximal width at the lower leg and at that location obtain axial images. T_1-weighted anatomical images were obtained with TR/TE=500/16 ms, slice thickness=5 mm, one slice, FOV=18x18 cm, matrix size= 256x256, N_{EX}=2. Diffusion weighted images were acquired in six different diffusion directions using: TR/TE= 4000/49 ms, slice thickness=5 mm, b=500 sec/mm^2, FOV 18x18 cm, matrix size=128x128, N_{EX}=4. The diffusion weighted images were registered to the unweighted image using an affine transformation. Then, the registered images were processed using the Philips PRIDE station fiber tracking tool to determine the tensor's eigenvalues (λ_1, λ_2 λ_3), and eigenvector (ε_1) and the ADC. For T_2 calculations, multiple spin-echo images were obtained with the same geometric parameters as the diffusion images and TR/TE=5000/20, 30, 40 and 60 ms; for T_2^* calculations gradient-echo EPI images were obtained with TR/TE 5000/20 and 40 ms. Custom MATLAB routines were used to process the images. An ROI was drawn around the Soleus muscle to determine the orientation of ε_1 with respect to a unit vector in the Z-direction. The mean values for ADC, λ_1, λ_2, λ_3, T_2 and T_2^* were also calculated in that same ROI.

Numerical Simulations: The effect of α on T_2 and T_2^* via the extravascular BOLD effect were predicted using the numerical model of Stables *et al* (2), assuming that muscle capillaries from a network of infinite cylinders, a blood volume fraction of 3%, B_0=3T, hematocrit (Hct)=0.4, oxyhemoglobin saturation=65%, a capillary radius of 5.4 μm, α of 26.9, 39.1, and 49 degrees (see below), a transverse diffusion coefficient of 1.3 x 10^{-3} mm^2 $\cdot$s^{-1} (see below) and calculating a blood-tissue magnetic susceptibility difference by adapting Eq. 1 of Spees *et al* (4). to also account for Hct and the effects of myoglobin.

Statistics: Means and standard deviations were calculated using SPSS 14. The general linear model with repeated measures was used to test for significant differences in α, ADC, λ_1, λ_2, λ_3, T_2 and T_2^*at the three different knee angles.

Results

Varying the degree of knee flexion from 0° to 10° of knee flexion significantly modified the orientation of the muscle fibers in the soleus muscle from 26° at complete extension to 49° at 10° degrees. There was no significant difference in muscle fiber orientation between complete extension and 5° knee flexion. Neither of the eigenvalues, ADC or the relaxation parameters T_2 and T_2^* were significantly modified at the different muscle fibers orientations (Table 1). Computer simulations predicted this response for the same degree of change in α.

Table 1. Relaxation and diffusion parameters calculated from DT, SE and GE images. Units for angles, °; diffusion parameters, 10^{-3} mm^2 $\cdot$s^{-1}; relaxation parameters, ms. *significant difference from knee angle = 0°. Mean ± SD is given.

Knee angle	α	λ_1	λ_2	λ_3	ADC	Predicted ΔT_2 vs. α=26.9	Measured T_2	Predicted ΔT_2* vs. α=26.9	Measured T_2*
0	26.9 ± 7.9	1.98 ± 0.04	1.64 ± 0.08	1.32 ± 0.01	1.64 ± 0.02		39.2 ± 2.7		23.8 ± 0.2
5	39.1 ± 9.0	1.99 ± 0.06	1.62 ± 0.08	1.30 ± 0.07	1.65 ± 0.04	-0.1	39.3 ± 2.5	-0.04	24.1 ± 0.1
10	49.0* ± 12.0	1.96 ± 0.05	1.59 ± 0.05	1.34 ± 0.09	1.62 ± 0.05	-0.2	40.2 ± 1.3	-0.1	23.8 ± 0.2

Conclusions

Changes in α from 27 to 50 in resting skeletal muscle do not significantly affect T_2 and T_2* at 3T. Measurements of T_2 during exercise at this field strength are not likely to be confounded by changes in muscle or capillary fiber orientations.

References

1. Damon BM, Gore JC: Biophysical basis of magnetic resonance imaging of small animals. Methods Enzymol 2004; 385: 19-40.
2. Stables LA, Kennan RP, Gore JC: Asymmetric spin-echo imaging of magnetically inhomogeneous systems: theory, experiment, and numerical studies. Magn Reson Med 1998; 40(3): 432-42.
3. Kawakami Y, Ichinose Y, Fukunaga T: Architectural and functional features of human triceps surae muscles during contraction. J Appl Physiol 1998; 85(2): 398-404.
4. Spees WM, Yablonskiy DA, Oswood MC, Ackerman JJ: Water proton MR properties of human blood at 1.5 Tesla: magnetic susceptibility, T(1), T(2), T*(2), and non-Lorentzian signal behavior. Magn Reson Med 2001; 45(4): 533-42.

Acknowledgments

NIH/NIAMS AR050101 and NIH/NCRR M01 RR 00095