Mapping Cross-Sectional Skeletal Muscle Asymmetry via High Angular Resolution Diffusion Imaging

D. C. Karampinos[1,2], K. F. King[3], D. Chen[1], B. P. Sutton[2,4], and J. G. Georgiadis[1,2]

[1]Mechanical Science and Engineering Department, University of Illinois at Urbana-Champaign, Urbana, IL, United States, [2]Beckman Institute, University of Illinois at Urbana-Champaign, Urbana, IL, United States, [3]GE Healthcare, Waukesha, WI, United States, [4]Bioengineering Department, University of Illinois at Urbana-Champaign, Urbana, IL, United States

Introduction:

The measured difference between the secondary and tertiary eigenvalues of the diffusion tensor implies asymmetry of cross-sectional (i.e. normal to the local fiber direction) diffusion in skeletal muscle, but its connection to muscle histoarchitecture or physiology explanation remains an open research question. Previous models of muscle diffusion have assigned the diffusion eigenvalues to distinct structural compartments [1], or have connected the observed cross-sectional asymmetry to the elliptical shape of the myofiber cross-section [2]. Experimental studies have also probed the origin of the difference in the two lower eigenvalues of the diffusion tensor by monitoring the variation of the eigenvalues while performing extension and contraction experiments [3-5]. All previous studies have relied on a low number of diffusion encoding directions (between 6 and 30) and the employment of the standard diffusion tensor model, which is only a second order approximation to a potentially more complex diffusion process. In the present work, we study the asymmetry of the diffusion coefficient on the transverse plane by performing a high angular resolution diffusion imaging (HARDI) acquisition and analyzing the results with a high order diffusion model. Our goal is to establish the existence of the asymmetry of transverse diffusion in measurements with a high number of diffusion encoding directions, and to examine the adequacy of the diffusion tensor model to describe the underlying diffusion process in the typical range of *b* values (500-600 s/mm^2) employed in in-vivo muscle DTI.

Materials and Methods:

Model formulation: The diffusion tensor model is first used to compute the diffusion tensor eigenvalues (λ_1, λ_2, λ_3) and eigenspace (spanned by the eigenvectors V_1, V_2, V_3). Then, we buid a a high order model by assuming that the principal direction of diffusion remains parallel to the main axis of the myofibers (i.e. the direction of the primary eigenvector) and introducing a more sophisticated representation of diffusion on the transverse plane defined by the secondary and tertiary eigenvectors. This representation is based on a circular spectral decomposition of the transverse diffusivity, in the spirit of the decomposition previously employed in mapping of intravoxel white matter fiber structures [6]. The ADC profile can then be decomposed as follows :

$$ADC(\theta,\varphi) = W_{\text{par}}\cos^2\theta + \sum_{k=0}^{N-1} W_{\text{tr}}(k)\cos(2k\varphi)\sin^2\theta$$

where θ and φ are the zenith and azimuth angle of the diffusion encoding in the system of coordinates defined by the tensor eigenvectors. The coefficients W_{par} and $W_{\text{tr}}(k)$ are computed by fitting the above expression with the sampled data. A matrix problem is formulated by using the linear least squares solution with additional regularization penalizing high order circular terms.

In vivo measurements: A single-channel lower extremity coil was used to scan the calf region of one male subject's right leg on a 3T full-body GE scanner. Diffusion-weighted images were acquired using a single-shot diffusion-weighted stimulated-echo EPI sequence with the following parameters: TR/TE=2000/52 ms, FOV=20x20 cm^2, slice thickness=10 mm, acquisition matrix=64x40 (5/8 partial phase encoding), and single average. Diffusion-weighted gradients were applied along 120 non-collinear directions with a *b*-value of 541 s mm^{-2}. The diffusion-encoding parameters were: δ=15 ms, Δ=40 ms, g=30 mT/m. Seven slices were acquired centered on the widest cross section of the calf muscle and fat suppression was performed using a spatial-spectral RF pulse.

Results and Discussion:

The mean diffusivity (MD=(λ_1+λ_2+λ_3)/3) of the diffusion tensor (Fig. 1b), and the first two coefficients of the circular decomposition (Figs. 1d and 1e) do not show any important spatial variation among different muscle regions. However, Fig. 1c, which is the map of the planar index CP of the diffusion tensor (where CP=2(λ_2-λ_3) / (λ_1+λ_2+λ_3)), shows significant variation of the transverse asymmetry in different regions corresponding to different muscle groups. The CP map establishes the presence of the asymmetry in the HARDI dataset and indicates that are spatial patterns in the diffusion asymmetry on the transverse plane.

The map of the secondary eigenvector (Fig. 1a) derived by employing the standard diffusion tensor analysis of the HARDI dataset reveals a coherent muscle organization, which is consistent with calf muscle anatomy, as it has been found previously in measurements using a lower number (thirty) of diffusion encoding directions [2]. Since the relatively coherent orientation of the secondary tensor eigenvector is a secondary effect relying on small diffusivity differences, a high number of diffusion encoding directions enables a more robust prediction of the orientation of the diffusion asymmetry on the transverse plane.

The coefficients of the circular spectrum for *k*>1 are close to zero and do not show any coherent spatial variation among different muscle regions. However, the map of the third coefficient of the circular spectrum (Fig. 1f) agrees with the tensor CP map, verifying the adequacy of the diffusion tensor model to describe the asymmetry of the transverse diffusivity at the employed *b*-value. Attention should be paid in generalizing the above conclusions at higher b-values, where the effect of secondary intravoxel fiber structures might be important.

We also note in passing the elevated values of CP and the $W_{\text{tr}}(1)$ in the periphery of lateral and medial gastrocnemius, which can be potentially connected with the adaptation of the muscle to increased lateral stress transmission in these regions [7].

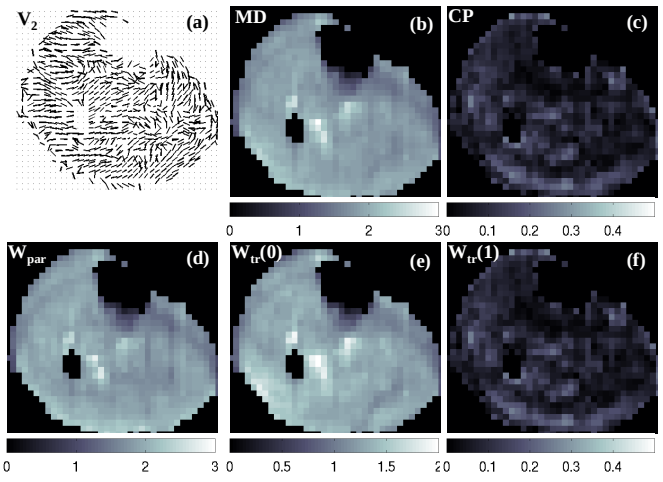

Fig. 1: analysis of HARDI data from a single slice: (a) Map of the projection of the secondary diffusion tensor eigenvector on the slice plane, (b) tensor mean diffusivity map, (c) tensor CP map, (d-f) maps of the circular decomposition coefficients: (d) for W_{par}, (e) for $W_{\text{tr}}(0)$, and (f) for $W_{\text{tr}}(1)$. (Fat ring shifted towards the posterior direction due to insufficient suppression of certain fat peaks).

Conclusion: Our high angular resolution diffusion imaging experiment on human calf muscle verified the asymmetry of the skeletal muscle diffusion on the transverse plane and indicated that the orientation of this asymmetry, characterized by the secondary eigenvector, is spatially coherent and consistent with muscle anatomy. A high order diffusion model accounting for a generalized diffusion process on the transverse plane showed that the diffusion tensor model is adequate for the description of the transverse diffusion profile at b=541 s/mm^2.

References: [1] Galban C. et al, Eur. J. Appl. Physiol. 93: 253-262, 2004, [2] Karampinos D. C. et al., ISMRM 2008, p. 2590, [3] Heemskerk A. et al., ISMRM 2008, p. 1787, [4] Hatakenaka M., J Magn Reson Imag 27: 932-937, 2008, [5] Deux J. F. et al., Eur. Radiol.18: 2303-2310, 2008, [6] Zhan W. et al., Magn Reson Med 49, 1077-1088, 2003, [7] van Donkelaar C. C. et al., J. Biomechanics 32: 755-762 ,1999.