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***In-vivo* Diffusion Tensor Imaging of Thoracic and Cervical Rat Spinal Cord at 7T**

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Abstract

In vivo diffusion tensor imaging (DTI) of rat cervical and thoracic spinal cord was performed using a three element phased array coil at 7T. The magnetic field was shimmed over the spinal cord in real time using an in-house developed automatic algorithm. Echo planar imaging (EPI) based diffusion weighted images (DWI) were acquired with 21 gradient encoding directions. The DWI images were tensor encoded and diffusion tensor metrics, fractional anisotropy (FA), mean diffusivity (MD), and longitudinal diffusivity (λ_p) and transverse diffusivity ($\lambda_{\perp}$), were determined for both white matter (WM) and gray matter (GM). The results on six normal rats indicated no significant differences in the diffusion tensor metrics between thoracic and cervical regions. However, the DTI-derived metrics in cervical spinal cord from our study are somewhat different from the published results in rats. The possible reasons for these differences are suggested.

Keywords

MRI; DTI; rat; spinal cord; thoracic cord; cervical cord; phased array coil; auto-shimming; anisotropic measures

Introduction

Detection and characterization of spinal cord injury (SCI) and other spinal cord pathologies require systematic studies that are possible only in animal models. For a number of reasons that include ease of handling, rodents are most commonly used in experimental SCI. Hitherto, SCI is most commonly characterized based on histology and biochemical evaluation of excised cord tissue. This requires sacrificing a large number of animals at different time points. More importantly, this can not be done in humans. A noninvasive technique that can detect and characterize SCI would allow us to follow the progression and/or regression of injury in the same group of animals and could be translated to humans for an objective evaluation of the efficacy of therapeutic intervention.

MRI is an excellent modality for non-invasively following pathological changes in spinal cord injury and other spinal cord diseases [1]. Diffusion tensor imaging (DTI) is a more recent MRI-based technique that has been shown to have the potential to detect subtle pathology that is not

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evident on conventional MRI [2,3]. DTI studies of spinal cord would be particularly useful in probing the integrity of white matter fiber tracts in traumatic spinal cord injury (SCI) [4].

Experimental SCI studies generally involve injury to the thoracic section of the spinal cord. In traumatic SCI, following the primary injury at the site of impact, secondary injury spreads along the length of the cord with time as a result of a series of biological events [5,6]. There is some evidence that suggests that injury spreads preferentially in the rostral direction [7]. Therefore, it is important to characterize with DTI large segments of spinal cord that include both thoracic and cervical regions at the same time. Prior to the application of DTI for investigating pathologic changes in injured spinal cord, it is essential to establish the normative DTI metrics in normal animals.

High signal-to-noise ratio (SNR) is important for in vivo DTI of rodent spinal cord. To obtain high SNR, implantable coils [8–11], single element external surface coil [12,13], and small birdcage volume coil [14] are used. Some of the disadvantages with implantable coils include: 1) invasive nature of implantation, 2) loss of SNR in longitudinal studies due to change in the resonant properties of the implanted coil with time, and 3) difficulty in implanting coil in the deeper cervical region of the spinal cord. Single element surface coil can produce high SNR from superficial segments of the spinal cord (T10–T12 sections of the thoracic segment). However, they provide limited spatial coverage and often produce poor SNR from the cervical segments [15], which are located deeper than the thoracic segments. A volume coil can offer uniform SNR through out the image at the expense of lower SNR compared to a surface coil. In contrast to a single surface coil or an implantable coil, phased array coils [16,17] offer high SNR and increased spatial coverage for imaging both thoracic and cervical regions. In addition, phased array coils offer parallel imaging capabilities [18–21].

Majority of the in-vivo DTI studies have so far been limited to thoracic region of rat [8,9,11, 12] and mouse spinal cord [13]. We are aware of only two DTI studies of cervical regions of spinal cord in rodents: one in rat [12] and the other one in mouse [14]. The study in [12], acquired DTI from both thoracic and cervical regions of rat spinal cord, used a single element RF coil, and acquired spin – echo based diffusion-weighted images (DWI) in six directions. These studies have reported statistically significant differences in the DTI-derived metrics between the thoracic and cervical sections of the spinal cord. The study in [12] has reported a relatively high FA value of 0.589 ± 0.008 in rat spinal cord GM. In contrast, the FA value of GM in mouse cervical cord [14] was reported to be 0.27 ± 0.01 . The high FA value in GM and large difference in the FA values of GM in the cervical cord between rat and mouse are somewhat unexpected.

The majority of the DTI studies in rodent spinal cord have been performed using the spin echo-sequence that involves long acquisition times. To reduce the scan time in spin echo-based DTI, either the total number of averages or the total number of diffusion encoding directions is often limited. This can limit the overall SNR that may overestimate the FA value and also produce bias in the DTI-derived metrics by the use of diffusion encoding schemes that may not be necessarily balanced [11]. In contrast to spin echo imaging, EPI can significantly reduce the total imaging time and allow the acquisition of DWI with relatively large number of diffusion encoding directions for reduced bias. However, EPI at high fields suffers from B_0 field inhomogeneities and eddy currents that produce geometric distortions and ghosting artifacts [9,22,23]. EPI based DTI studies were reported with high quality shimming and proper pre-emphasis adjustments [8,9,11,14]. As indicated earlier, these studies used implantable coils for high SNR and were limited to the thoracic region.

In this study, we focused on the rat spinal cord. To the best of our knowledge, these are the first EPI-based DTI studies of both thoracic and cervical regions of rat spinal cord. We believe that

the use of balanced icosahedral diffusion encoding scheme ICOSA21 [24,25] with relatively large number of gradient encoding directions, along with phased array coil for improved SNR and spatial coverage, has resulted in robust and unbiased estimation of the DTI metrics. These aspects distinguish our studies from those previously published.

Materials and Methods

All MRI studies were performed on a 7T Bruker 70/30 USR scanner (Bruker Biospec, Karlsruhe, Germany) with BGA-20/BS-30 gradient/shim system. The maximum gradient amplitude was 200 mT/m with a slew rate of 640 T/m/s. A three element phased array coil, described below, was used in the receive-only mode and a 154 mm diameter volume coil (linear birdcage resonator, provided by the MR scanner manufacturer) was used for transmitting the RF power. All images were acquired using Bruker's PARAVISION 4.0 software.

Male Sprague-Dawley rats, weighing between 300 and 350 g, were used in these studies. The animals were placed in the supine position and were maintained under anesthesia during the entire MRI scan with a mixture of 2.5% isoflurane, air, and oxygen, administered through a Harvard rodent ventilator (Model 683, Harvard Apparatus, South Natic, MA). The animals' body temperature was maintained at 37 °C using the air heating system (Model 11007B, SA Instruments, Inc., Stonybrook, NY). Respiration and rectal temperature were monitored throughout the scan with an MR compatible physiologic monitoring unit (Model 1025, SA Instruments, Inc., Stonybrook, NY). A Nonin pulse oximeter (Model 8600V, NONIN, Plymouth, MN) was used to monitor the heart rate and oxygen levels.

RF coil design

A three element phased array coil was designed for imaging both thoracic and cervical regions of the rat spinal cord [15]. Briefly, each element in the array has 20 mm × 50 mm dimensions on a planar surface. With the coils wrapped around a 57 mm OD × 51 mm ID acrylic tube, the measured self-inductance was 89 nH and the loaded resistance was 4.9Ω. By judiciously choosing the dimensions of each single element, we have demonstrated that high resolution images can be obtained from both thoracic and cervical regions [15]. Coupling between the nearest neighbors was minimized by overlapping the coil elements and a simple capacitive network was used for decoupling the next neighbor elements. After connecting all the electronics, including tuning, matching, decoupling and balun electronics, the measured isolation between the neighboring and next neighboring coils was -15 dB and -22 dB, respectively.

Anatomical MRI

A rapid acquisition with relaxation enhancement (RARE) sequence with the following parameters was used to acquire anatomical images: TR/TE = 1835.9/23.8 ms, ETL = 4, NEX = 8, slice thickness = 1 mm, matrix dimensions = 256 × 256 and field-of-view (FOV) = 4.5 cm × 4.5 cm, where TR and TE are the repetition and echo times, respectively, ETL is the echo train length, and NEX is the number of averages. With these parameters, the in plane resolution was 0.17 mm × 0.17 mm and total imaging time was 15 min 21sec. Sum-of-squares (SOS) technique was used to reconstruct composite images from the individual coil images.

EPI-DTI Protocol

The DWI images were acquired with a navigator echo based spin-echo EPI sequence. The imaging parameters were TR/TE = 1500.00/30.37 ms, number of shots = 8, effective bandwidth = 200 kHz, matrix dimensions = 128 × 128, number of contiguous slices = 5 and the total imaging time = 18 min 24 sec. The PARAVISION 4.0 software used in this study does not allow multi-channel acquisition of the EPI data. To demonstrate the feasibility of DTI

characterization using phased array coils in the current study, the EPI data were acquired from one coil at a time and post processed to reconstruct SOS images. The phased array coil produced lower SNR in the cervical region compared to the thoracic region. To compensate for the lower SNR in cervical region, a higher field-of-view FOV and slice thickness (square FOV = 4 cm and slice thickness = 2 mm) were selected for the cervical region and smaller FOV and slice thickness (square FOV = 3 cm and slice thickness = 1 mm) were selected for the thoracic region. To minimize the eddy current related ghosting artifacts, optimal values for pre-emphasis adjustments were performed manually and these values were stored in the pre-emphasis file for acquiring the EPI data.

The DTI acquisition parameters were, diffusion gradient duration = 4 ms, diffusion gradient separation = 15.04 ms, number of diffusion directions = 21, diffusion experiments per direction = 4, number of b_0 images (without diffusion gradient) = 8 and b value = 500 s/mm². The gradient directions were derived from the ICOSA21 encoding scheme [24]. The DWI images acquired from each coil were combined using the SOS reconstruction technique and the images were tensor encoded using MedINRIA software [26].

To measure the DTI statistics from WM and GM, ROI's were drawn on slices from T12-T7 region of the thoracic spinal cord and C3-C7 region of the cervical spinal cord.

The value of FA is sensitive to the underlying SNR and higher FA values result from a water phantom for lower SNR [11]. Ideally, the FA value of water should be zero. However, low SNR and other factors result in FA value that is non-zero. Hence, the FA values determined in water phantom can be used as a good quality control. In order to investigate the effects of SNR on the FA values in our study, DWI images were acquired with different number of averages. The SNR inside the phantom was measured at a depth of ~17 mm from the surface of the coil. This depth approximately corresponds to the location of the cervical cord from the surface.

Field Map Shimming

The bony structures surrounding the spinal cord create a harsh magnetic environment locally. Thus homogenizing the magnetic field over spinal cord is very important for high quality EPI. Because of the curvature of the spinal cord, the manufacturer supplied shimming routines do not work well for spinal cord imaging. Therefore, we have developed and implemented an automatic and real time shimming technique. This field-map based technique allows shimming in real time over any arbitrary shaped geometry and size. Before the acquisition of DWI images, the spinal cord regions were shimmed using this in-house developed shimming technique. The shimming was performed over all the 14 shims incorporated into our shim coil. Although a number of different ways of implementing the field-map shimming technique were proposed [27–29], all these techniques are similar in three fundamental ways: 1) acquiring B_0 field-maps in 3D, 2) solving a system of linear equations to estimate the shim currents for in-vivo shimming, and 3) implementing the technique in real time. Brief description about the implementation of the field-map shimming technique in current studies is described below.

Before the shimming technique was applied for in-vivo studies, a reference (or calibration) field-map data was acquired from a water phantom with 5 cm × 5 cm × 5 cm dimensions. The reference maps were acquired using a 3D multiple gradient echo (MGE) sequence. Both magnitude and phase data at two different echo times were acquired. The imaging parameters were TR=100 ms, TE1/TE2= 2.38 ms/4.76 ms, square FOV= 5.5 cm, matrix dimensions = 64 × 64 × 32 and number of averages = 4. The phase images at the two echo times were subtracted and the resulting images were unwrapped [30] to generate the reference data for each of the 14 shims.

For in-vivo shimming, images were acquired with the 3D MGE sequence with the following acquisition parameters: TR=100 ms, TE1/TE2= 2.38/4.76 ms, square FOV= 4.5 cm, matrix dimensions = $64 \times 64 \times 32$ and number of averages = 1.

The field-maps were loaded into the software that was developed under IDL 6.3 (ITT Visual Solutions, Boulder CO) programming language. The software allows the user to specify an ROI of any shape and size for shimming. Once the user finishes drawing the ROI, the software solves a system of linear equations and calculates new shim currents for shimming over the ROI. The software is integrated into PARAVISION using a macro and works in real time.

All SNR measurements were performed on the SOS b_0 images. The signal was measured as the mean signal intensity of the pixels within a ROI in the image and the noise was measured as the standard deviation (SD) of the pixels in an artifact-free background.

Results

The FA values in water phantom as a function of the SNR in the water phantom are shown in Fig. 1. Higher SNR resulted in lower mean and SD for FA. For the total number of DWI of 92 (21 gradient encoding directions $\times$ 4 averages + $8 \times b_0$), the value used in our spinal cord studies, the estimated value of FA from water phantom was 0.05 ± 0.005 . Based on the regression analysis, the FA value linearly decreases with SNR.

To evaluate the in-vivo performance of the shimming technique, EPI-based diffusion weighted images were acquired from a rat spinal cord both with global shimming, provided by the manufacturer, and our field-map shimming. These images are shown in Fig. 2. It can be observed from this figure that the spinal cord geometry, particularly the H-shaped gray matter structure, is much better preserved with our field-map shimming than the global –shimming, implemented by the manufacturer.

Fig. 3 shows the b_0 images from different coils and the SOS reconstructed image. The absence of ghosting related artifacts can be easily appreciated on this figure. A comparison of the EPI image (Fig. 3(d)) with the RARE image (Fig. 3(e)) also shows that the EPI image is relatively free of geometric distortions. The measured SNR values for the thoracic and cervical regions are 92 ± 16.1 and 97 ± 18.5 , respectively.

The FA and MD maps at the thoracic and cervical regions are shown in Fig. 4. In addition to the FA and MD maps, the transverse (TD or $\lambda_{\perp}$) and longitudinal (LD or $\lambda_{\parallel}$) diffusivities were also generated and the quantitative values of various DTI-derived metrics are summarized in Table 1 and Fig. 5. Two-tailed student t-tests have indicated no statistical difference between the thoracic and cervical region DTI values ($p \geq 0.1$).

Discussion

We have described the DTI studies of rat spinal cord in the thoracic and cervical regions. As pointed out earlier, a number of DTI studies of rodent spinal cord have been reported. However, in contrast to the previous studies that used either implantable coils or single element coils to acquire the DWI data, we have used a three element phased array coil for high SNR with increased spatial coverage. With careful shimming over the spinal cord and pre-emphasis adjustments, the EPI-based DWI data was acquired with high SNR, minimal geometric distortions and ghosting artifacts. The DWI data was acquired using a balanced encoding scheme with 21 directions for estimating the DTI metrics with reduced bias. The version 4.0 of the PARAVISION software used in this study doesn't allow multi-channel acquisition of the EPI images. To demonstrate the feasibility of DTI characterization using phased array coils in the current study, the EPI images were acquired from one coil at a time and post processed

to reconstruct SOS images. This has increased the imaging time three times. Otherwise, it is possible to acquire higher resolution images at reasonable acquisition times.

To the best of our knowledge only one *in-vivo* study was reported on DTI from both thoracic and cervical regions [12] of rat spinal cord. Our FA values of 0.38 ± 0.039 (thoracic) and 0.36 ± 0.044 (cervical) in GM were significantly different from those reported in [12]. We believe that this disagreement comes from the differences in the methods. In contrast to our study, the study in [12] was performed using six diffusion encoding directions and a single surface coil which might have different SNR than our phased array coil. The study in [12] has reported FA values of 0.162 ± 0.003 and 0.168 ± 0.003 from water phantom in two different scan sessions. In contrast, our FA values from water phantom using the ICOSA21 encoding scheme is 0.05 ± 0.005 . Based on our small FA values from the water phantom, we believe that the FA values determined in our study are robust. Also, the FA values of GM in [12] are also significantly different from the reported values from the mouse spinal cord [14]. Our FA values in GM are much closer to the values in [14] for mouse cervical spinal cord. There is little basis to expect significant differences in the DTI metrics between rat and mouse spinal cord.

In summary, we described an EPI based DTI characterization of rat spinal cord in both thoracic and cervical regions using a phased array coil and filed-map based shimming. Our results indicated no statistical difference in the DTI characteristics of either GM or WM between thoracic and cervical regions.

Abbreviations

DTI	diffusion tensor imaging
EPI	echo planar imaging
FA	fractional anisotropy
MD	mean diffusivity
TD	transverse diffusivity
LD	longitudinal diffusivity
WM	white matter
GM	gray matter
SCI	spinal cord injury
SNR	signal-to-noise ratio
DWI	diffusion-weighted images

SD

standard deviation

RARE

rapid acquisition with relaxation enhancement

SOS

sum-of-squares

References

1. Narayana, PA. Magnetic resonance imaging in spinal cord injury. In: Marwah, J.; Dixon, E.; Banik, N., editors. Handbook of CNS trauma. Scottsdale, AZ: Prominent Press; 2000.
2. DeBoy CA, Zhang J, Dike S, Shats I, Jones M, Reich DS, Mori S, Nguyen T, Rothstein B, Miller RH, Griffin JT, Kerr DA, Calabresi PA. High resolution diffusion tensor imaging of axonal damage in focal inflammatory and demyelinating lesions in rat spinal cord. *Brain* 2007;130(Pt 8):2199–2210. [PubMed: 17557778]
3. Deo AA, Grill RJ, Hasan KM, Narayana PA. In vivo serial diffusion tensor imaging of experimental spinal cord injury. *J Neurosci Res* 2006;83(5):801–810. [PubMed: 16456864]
4. Herrera JJ, Chacko T, Narayana PA. Histological correlation of diffusion tensor imaging metrics in experimental spinal cord injury. *J Neurosci Res* 2008;86(2):443–447. [PubMed: 17868152]
5. Hagg T, Oudega M. Degenerative and spontaneous regenerative processes after spinal cord injury. *J Neurotrauma* 2006;23(3–4):264–280. [PubMed: 16629615]
6. Moreno-Flores MT, Avila J. The quest to repair the damaged spinal cord. *Recent Patents CNS Drug Discov* 2006;1(1):55–63.
7. Massey JM, Amps J, Viapiano MS, Matthews RT, Wagoner MR, Whitaker CM, Alilain W, Yonkof AL, Khalyfa A, Cooper NG, Silver J, Onifer SM. Increased chondroitin sulfate proteoglycan expression in denervated brainstem targets following spinal cord injury creates a barrier to axonal regeneration overcome by chondroitinase ABC and neurotrophin-3. *Exp Neurol* 2008;209(2):426–445. [PubMed: 17540369]
8. Bilgen M, Elshafiey I, Narayana PA. In vivo magnetic resonance microscopy of rat spinal cord at 7 T using implantable RF coils. *Magn Reson Med* 2001;46(6):1250–1253. [PubMed: 11746595]
9. Fenykes DA, Narayana PA. In vivo diffusion tensor imaging of rat spinal cord with echo planar imaging. *Magn Reson Med* 1999;42(2):300–306. [PubMed: 10440955]
10. Ford JC, Hackney DB, Joseph PM, Phelan M, Alsop DC, Tabor SL, Hand CM, Markowitz RS, Black P. A method for in vivo high resolution MRI of rat spinal cord injury. *Magn Reson Med* 1994;31(2):218–223. [PubMed: 8133758]
11. Madi S, Hasan KM, Narayana PA. Diffusion tensor imaging of in vivo and excised rat spinal cord at 7 T with an icosahedral encoding scheme. *Magn Reson Med* 2005;53(1):118–125. [PubMed: 15690510]
12. Ellingson BM, Kurpad SN, Li SJ, Schmit BD. In vivo diffusion tensor imaging of the rat spinal cord at 9.4T. *J Magn Reson Imaging* 2008;27(3):634–642. [PubMed: 18224673]
13. Kim JH, Loy DN, Liang HF, Trinkaus K, Schmidt RE, Song SK. Noninvasive diffusion tensor imaging of evolving white matter pathology in a mouse model of acute spinal cord injury. *Magn Reson Med* 2007;58(2):253–260. [PubMed: 17654597]
14. Callot V, Duhamel G, Cozzone PJ, Kober F. Short-scan-time multi-slice diffusion MRI of the mouse cervical spinal cord using echo planar imaging. *NMR Biomed* 2008;21(8):868–877. [PubMed: 18574855]
15. Kishore MV, Bankson JA, Narayana PA. Three Element Phased Array Coil for Imaging of Rat Spinal Cord at 7T. *Magn Reson Med* 2008 Dec;60(6):1498–505. [PubMed: 19025892]
16. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn Reson Med* 1990;16(2):192–225. [PubMed: 2266841]
17. Wright SM, Wald LL. Theory and application of array coils in MR spectroscopy. *NMR Biomed* 1997;10(8):394–410. [PubMed: 9542737]

18. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med* 2002;47(6):1202–1210. [PubMed: 12111967]
19. Kyriakos WE, Panych LP, Kacher DF, Westin CF, Bao SM, Mulkern RV, Jolesz FA. Sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE RIP). *Magn Reson Med* 2000;44(2):301–308. [PubMed: 10918330]
20. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med* 1999;42(5):952–962. [PubMed: 10542355]
21. Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays. *Magn Reson Med* 1997;38(4):591–603. [PubMed: 9324327]
22. Reeder SB, Atalar E, Bolster BD Jr, McVeigh ER. Quantification and reduction of ghosting artifacts in interleaved echo-planar imaging. *Magn Reson Med* 1997;38(3):429–439. [PubMed: 9339445]
23. Schmuthorst VJ, Dardzinski BJ, Holland SK. Simultaneous correction of ghost and geometric distortion artifacts in EPI using a multiecho reference scan. *IEEE Trans Med Imaging* 2001;20(6):535–539. [PubMed: 11437113]
24. Hasan KM, Parker DL, Alexander AL. Comparison of gradient encoding schemes for diffusion-tensor MRI. *J Magn Reson Imaging* 2001;13(5):769–780. [PubMed: 11329200]
25. Hasan KM, Narayana PA. Computation of the fractional anisotropy and mean diffusivity maps without tensor decoding and diagonalization: Theoretical analysis and validation. *Magn Reson Med* 2003;50(3):589–598. [PubMed: 12939767]
26. Fillard, P.; Toussaint, N.; Souplet, JC.; Vercauteren, T. MedINRIA Software. 1.7.1. France: INRIA Sophia Antipolis; 2008.
27. Kim DH, Adalsteinsson E, Glover GH, Spielman DM. Regularized higher-order in vivo shimming. *Magn Reson Med* 2002;48(4):715–722. [PubMed: 12353290]
28. Schneider E, Glover G. Rapid in vivo proton shimming. *Magn Reson Med* 1991;18(2):335–347. [PubMed: 2046515]
29. Webb P, Macovski A. Rapid, fully automatic, arbitrary-volume in vivo shimming. *Magn Reson Med* 1991;20(1):113–122. [PubMed: 1943653]
30. Jenkinson M. Fast, automated, N-dimensional phase-unwrapping algorithm. *Magn Reson Med* 2003;49(1):193–197. [PubMed: 12509838]

Figure 1. FA values in a water phantom as a function of SNR

The FA values were reported as mean $\pm$ SD. The linear regression equation is shown in the top.

Figure 2. In-vivo rat spinal cord images

(a) Anatomical RARE image, (b) EPI based diffusion weighted image with global shimming, and (c) image with field-map shimming

Figure 3. Individual and SOS images from spinal cord

b_0 images from each element of the phased coil are shown in (a), (b), and (c). The composite SOS image is shown in (d). For reference, the anatomical image is shown in (e).

Figure 4. FA and ADC maps from thoracic and cervical regions

The RARE image from a thoracic spinal cord (T12-T10) is shown in (a). The corresponding FA and ADC maps are shown in (b) and (c), respectively. The RARE image from a cervical spinal cord (C3-C5) is shown in (d) and its corresponding FA and MD maps are shown in (e) and (f), respectively.

Figure 5. DTI metrics from both GM and WM in rat spinal cord

MD, TD (transverse) and LD (longitudinal) diffusivities are expressed as $10^{-3} \text{ mm}^2/\text{sec}$. The bars indicate the SD and the p -value are based on the two-tailed Student t-test.

Table 1
DTI values from thoracic and cervical spinal cord ($N = 6$), expressed as $mean \pm SD$. The units of MD, TD ($\lambda_{\perp}$) and LD (λ_p) are expressed as $10^{-3} \text{ mm}^2/\text{sec}$.

		FA	MD	$\lambda_{\perp}$	λ_p
Thoracic	WM	0.83 ± 0.052	0.86 ± 0.056	0.29 ± 0.069	1.94 ± 0.115
	GM	0.38 ± 0.039	1.04 ± 0.071	0.77 ± 0.116	1.49 ± 0.106
Cervical	WM	0.82 ± 0.052	0.93 ± 0.104	0.33 ± 0.053	2.15 ± 0.260
	GM	0.36 ± 0.044	1.13 ± 0.152	0.83 ± 0.058	1.54 ± 0.202