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The impact of post-processing on spinal cord diffusion tensor imaging

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ABSTRACT

Diffusion tensor imaging (DTI) provides information about the microstructure in the brain and spinal cord. While new neuroimaging techniques have significantly advanced the accuracy and sensitivity of DTI of the brain, the quality of spinal cord DTI data has improved less. This is in part due to the small size of the spinal cord (ca. 1 cm diameter) and more severe instrumental (e.g. eddy current) and physiological (e.g. cardiac pulsation) artefacts present in spinal cord DTI. So far, the improvements in image quality and resolution have resulted from cardiac gating and new acquisition approaches (e.g. reduced field-of-view techniques). The use of retrospective correction methods is not well established for spinal cord DTI. The aim of this paper is to develop an improved post-processing pipeline tailored for DTI data of the spinal cord with increased quality. For this purpose, we compared two eddy current and motion correction approaches using three-dimensional affine (3D-affine) and slice-wise registrations. We also introduced a new robust-tensor-fitting method that controls for whole-volume outliers. Although in general 3D-affine registration improves data quality, occasionally it can lead to misregistrations and biassed tensor estimates. The proposed robust tensor fitting reduced misregistration-related bias and yielded more reliable tensor estimates. Overall, the combination of slice-wise motion correction, eddy current correction, and robust tensor fitting yielded the best results. It increased the contrast-to-noise ratio (CNR) in FA maps by about 30% and reduced intra-subject variation in fractional anisotropy (FA) maps by 18%. The higher quality of FA maps allows for a better distinction between grey and white matter without increasing scan time and is compatible with any multi-directional DTI acquisition scheme.

Introduction

In the past years, more sophisticated imaging techniques such as functional (Eippert et al., 2009; Lotze et al., 2006; Sprenger et al., 2012; Wietek et al., 2008) and diffusion magnetic resonance imaging (MRI) (Agosta et al., 2007; Budde et al., 2007; Ciccarelli et al., 2007; Mulcahey et al., 2012) have become available for imaging the spinal cord. Diffusion MRI allows for non-invasive tracking of water diffusion (Le Bihan et al., 1986; Turner et al., 1990) and can be used to map brain anatomy (Bach et al., 2011; Basser et al., 1994; Draganski et al., 2011; Mohammadi et al., 2012b; Mueller et al., 2011; Pierpaoli and Basser, 1996). In clinical research diffusion tensor imaging (DTI), a particular implementation of diffusion MRI, has become a wide-spread and successful imaging method (Duning et al., 2009; Keller et al., 2011; Meinzer et al., 2010; Warnecke et al., 2010). For example, the scalar DTI-index denoted as fractional anisotropy (FA) has been reported to be sensitive to white matter integrity in health and disease in the brain (Deppe et al., 2007; Freund et al., 2012b; Pierpaoli et al., 2001) and spinal cord (Agosta et al., 2007; Budde et al., 2007; Ciccarelli et al., 2007; Freund et al., 2012c; Mulcahey et al., 2012).

The spinal cord is a small structure (ca. 1 cm in total diameter) and specific localization of injuries in the spinal cord requires a robust distinction between grey matter (GM) and white matter (WM) (Freund et al., 2012a). Up to now, most diagnostic studies in the spinal cord were limited by the quality and resolution of the DTI reconstruction (e.g. equal to or more than 1 mm^2 in-plane resolution (Agosta et al., 2007; Budde et al., 2007; Ciccarelli et al., 2007; Freund et al., 2011; Mulcahey et al., 2012; Roser et al., 2010)). Due to the cylindrical symmetry of the spinal cord, usually thick slices (about 5 mm) with maximal in-plane resolution are acquired leading to particularly long EPI readout times (Finsterbusch, 2009b, 2012; Rossi et al., 2008; Wilm et al., 2007, 2009) and making the signal susceptible to physiological and instrumental artefacts. Physiological artefacts caused by bulk motion of the cord and cerebrospinal fluid (CSF) pulsation can result in slice-to-slice displacement, deformation, and signal-loss due to a shift of the echo centre in k-space (Chung et al., 2010; Mohammadi et al., 2012a; Skare and Andersson, 2001). Instrumental artefacts caused by eddy currents (Haselgrove and Moore, 1996; Jezzard et al., 1998; Mohammadi et al., 2010), gradient inhomogeneities (Bammer et al., 2003; Mohammadi et al., 2012d; Nagy et al., 2007), vibration artefacts (Gallichan et al., 2010; Mohammadi et al., 2012c), and RF

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transmit field inhomogeneities (Lutti et al., 2010, 2012) can lead to image distortions (Mohammadi et al., 2010), affect the diffusion weighting (Mohammadi et al., 2012d), and perturb the signal intensity (Gallichan et al., 2010; Lutti et al., 2010, 2012; Mohammadi et al., 2012c). Up to now, the improvements in image quality and resolution were based on cardiac gating (Rossi et al., 2008; Wheeler-Kingshott et al., 2002a,b) and new acquisition technology, such as reduced field-of-view techniques (Finsterbusch, 2009b, 2012; Rossi et al., 2008; Wheeler-Kingshott et al., 2002a,b; Wilm et al., 2007, 2009), stronger diffusion weighting gradients (Wilm et al., 2009), increased number of averages (Rossi et al., 2008), and time-efficient monopolar diffusion-weighting schemes (Finsterbusch, 2009a; Morelli et al., 2010).

Surprisingly, the use of post-processing correction methods was rarely reported in spinal cord DTI (Barakat et al., 2012; Cohen-Adad et al., 2011; Freund et al., 2012c; Lundell et al., 2013; Wilm et al., 2009). However, using post-processing correction methods could potentially reduce remaining artefacts and even compensate for some of the drawbacks of the reported new acquisition approaches. For example, the methods that are related to improved diffusion weighting (stronger diffusion gradients or monopolar diffusion schemes) usually increase instrumental artefacts such as eddy currents (Haselgrove and Moore, 1996; Jezzard et al., 1998) and could benefit from retrospective eddy current correction (see, e.g., Wilm et al., 2009). Physiological artefacts in DTI affect data quality and can be reduced retrospectively using robust tensor fitting (Mangin et al., 2002; Walker et al., 2011; Zwiers, 2010) and linear modelling of artefacts (Mohammadi et al., 2012a). Increasing the number of averages might lead to more subject motion artefacts, which can be corrected using three-dimensional (3D) affine (e.g. Cohen-Adad et al., 2011; Mohammadi et al., 2010; Muñoz Maniega et al., 2007) or slice-wise (e.g. Mohammadi et al., 2010; Speck et al., 2006) registration methods.

The aim of this paper is to provide an improved processing pipeline for robust DTI in the spinal cord, which is compatible with previously suggested acquisition methods. To this end, we determine the effect of pre-processing (none, 3D-affine, and slice-wise eddy current and motion correction) and tensor estimation (ordinary least squares vs. robust tensor fitting) methods on the image quality and contrast-to-noise ratio (CNR) between GM and WM.

Methods

Subjects

Nine healthy adult volunteers (1 female, 8 males, age: 35 ± 8) participated in the study approved by the local ethics committee after giving written informed consent.

Data acquisition

Experiments were performed on a MAGNETOM Trio, a Tim System, 3T scanner (Siemens Healthcare, Erlangen, Germany) operated with an RF body transmit coil and a 12-channel (12-ch) receive-only head, 4-ch neck and 24-ch spine coil. Only the 4 neck channels and the 6 posterior head channels were used, since they provided full coverage of the scanned area. DTI data were acquired with a cardiac-gated monopolar diffusion sequence (Morelli et al., 2010) using the following parameters: 30 diffusion-weighted (DW) images ($b = 500$ s/mm^2), 5T2-weighted images without diffusion weighting ($b = 0$ images), 5 mm slice thickness, with 10% inter-slice gap, 10 slices perpendicularly oriented to the spine, 5/8 Partial-Fourier Imaging in phase-encoding direction, phase oversampling 50%, and a cardiac trigger delay of 200 ms. Two slightly different in-plane resolutions, field-of-view (FoV), and echo times (TEs) were used in this study: $176 \times 40/176 \times 60$ acquisition matrix, $123 \times 28/128 \times 43$ mm^2 FoV, $0.7 \times 0.7/0.73 \times 0.73$ mm^2 in-plane, echo time of TE = 73/75 ms, slice repetition time of TR = 290/350 ms. The gated data were acquired in blocks of two slices per cardiac cycle. The minimal time between successive triggers was 1800 ms. The reduced FoV was achieved using two saturation pulses (Heidemann et al., 2009) (see Fig. 1). Subjects S1–S6 and S8 were measured with the first set of parameters, and subjects S7 and S9 with the second set of parameters. The difference between the two protocols was small and we did not observe any difference in the resulting image quality. Each DTI dataset was acquired four times, resulting in 140 images for each subject. Altogether, this resulted in a total acquisition time of about 5.8 min (as estimated by the sequence simulator), but could be longer depending on the participant's heart rate. Subsequently, the abbreviations x, y, and z are used for the directions right-left (frequency encoding), anterior-posterior (phase encoding), and head-feed (slice selection), respectively.

Pre-processing and tensor estimation

First, the in-plane field-of-view was chopped to 28×28 mm^2 for each DTI dataset to exclude non-spine tissue. Next, the images were interpolated to a higher in-plane resolution of 0.35×0.35 mm^2. Finally, the data were corrected for motion and eddy current artefacts using three different registration methods: (a) none, (b) 3D-affine, and (c) combination of rigid-body and slice-wise motion correction (details are summarised in Table 1). The 3D-affine registration corrects for rigid-body subject motion and linear eddy current effects (see (Mohammadi et al., 2010)). Before applying the slice-wise registration, a 3D-affine registration was performed to reduce 3D translation in x- and y-direction as well as scaling effects in y-direction. We restricted the slice-wise registration to correct only for in-plane x- and y-translation as well as for in-plane scaling in y-direction, because we observed most variation in those directions. We did not correct for in-plane rotation and shearing effects, which were less pronounced and more difficult to estimate robustly.

After pre-processing, the FA was estimated using two different tensor-fitting methods: (a) ordinary least squares (Koay et al., 2006) and (b) a new robust-fitting method based on (Mohammadi et al., 2012a; Zwiers, 2010). We extended the robust-fitting method of Zwiers to account also for whole-volume outliers (e.g. due to 3D-affine

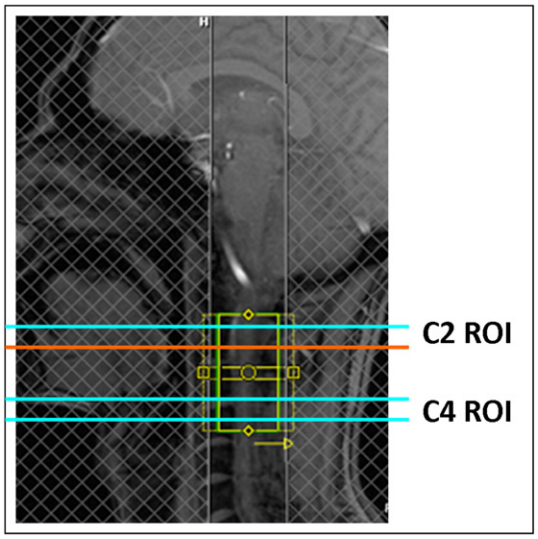

Fig. 1. Positioning of the field of view (small central solid green/yellow) covering cervical segments C2 and C4 (sagittal view). Reduced field of view was achieved by minimizing the phase-encoding steps in the anterior-posterior direction and avoiding consequential fold-over by two spatial saturation pulses (shaded regions). The slice positions of the grey and white matter region of interest (ROI) in the upper part of C2 (C2 ROI, cyan and orange horizontal lines) and in the lower part of C4 (C4 ROI, two cyan horizontal lines) are depicted. The grey and white matter ROIs at the position of the orange line are shown in Fig. 2.