Diffusion Tensor Imaging Enhanced Anisotropic MRE of the Brain

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Introduction

Magnetic Resonance Elastography (MRE) provides us with a method of non-invasively measuring the mechanical properties of the brain, and it has been used in previous studies *in vivo* to determine the viscoelasticity of brain tissues [1, 2]. Most MRE studies have assumed brain tissue isotropy, although some brain regions have been shown to be weakly anisotropic in rheological tests [3]. Diffusion tensor imaging (DTI) can be used to measure structural anisotropy in human brain [4]. Here, we combine MRE and DTI to investigate anisotropic viscoelasticity in the human brain *in vivo*.

Methods

Eight healthy volunteers (1 male, 7 females, 23 – 42 years of age) were scanned using a Philips Achieva 3TX scanner and an 8-channel head coil (Philips, Best, The Netherlands). MRE and DTI were acquired using a matrix of 64x64x7(3mm^3 isotropic voxels). MRE was performed as previously described [1] with 80 Hz mechanical vibration. DTI was acquired using a B-value of 800s/mm^2, 32 gradient directions, TR/TE =11.8s/83ms. The local coordinate system for MRE reconstruction was set to match the principal eigenvector from DTI. Shear moduli, $\mu_{\parallel}$ and $\mu_{\perp}$, were reconstructed, and the fractional anisotropy (FA) of DTI and the FA of shear moduli (μFA) computed. ROIs were drawn for grey matter (GM) and white matter (WM), and the mean $\mu_{\parallel}$, $\mu_{\perp}$, DTI FA and μFA were calculated for GM and WM. The WM voxels were further categorized into four groups with DTI FA values between 0.2-0.3, 0.3-0.4, 0.4-0.5, 0.5-0.6. The mean $\mu_{\parallel}$, $\mu_{\perp}$ and μFA were calculated for each group of voxels in each subject, and were then averaged over all subjects. ANOVA with post-hoc analysis was used to compare $\mu_{\parallel}$ and $\mu_{\perp}$, and μFA, and DTI FA across the DTI FA groups. ROIs (Table 1) for the white matter structures of internal capsule (IC), corpus callosum (CC) and corona radiata (CR) were drawn based on the DTI FA in these locations. The mean and standard errors of the $\mu_{\parallel}$, $\mu_{\perp}$, μFA and DTI FA in these WM structures were calculated over all subjects. ANOVA was used to compare $\mu_{\parallel}$ and $\mu_{\perp}$, and μFA, and DTI FA across these WM structures.

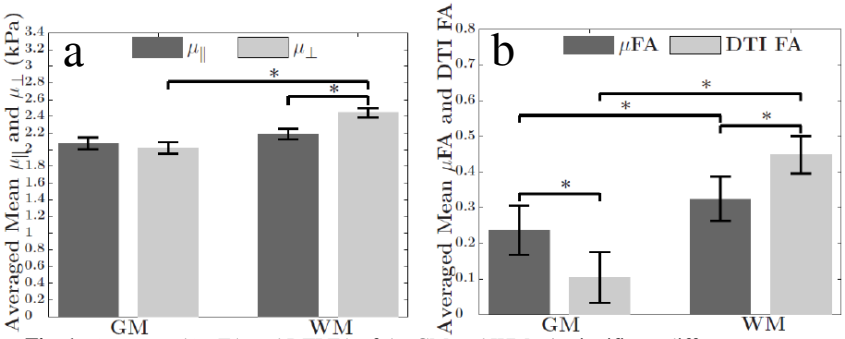

Fig. 1 (a) $\mu_{\parallel}$, $\mu_{\perp}$, (b) μFA and DTI FA of the GM and WM. (*- significant difference).

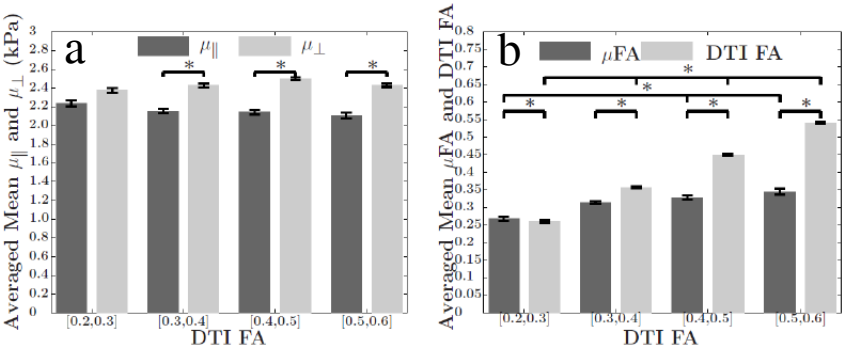

Fig. 2 (a) $\mu_{\parallel}$, $\mu_{\perp}$, (b) μFA and DTI FA of WM voxels with different DTI FA(*- significant difference).

	$\mu_{\parallel}$ (kPa)	$\mu_{\perp}$ (kPa)	μFA	DTI FA
IC	1.67 (0.05)	2.04 (0.04)	0.46 (0.02)	0.46 (0.00)
CC	2.18 (0.05)	2.32 (0.09)	0.37 (0.01)	0.55 (0.01)
CR	1.77 (0.06)	2.54 (0.04)	0.35 (0.01)	0.41 (0.00)

Table 1 $\mu_{\parallel}$, $\mu_{\perp}$, μFA and DTI FA at IC, CC and CR with standard errors in parentheses.

Results and Discussion

In WM, $\mu_{\perp}$ was significantly greater than $\mu_{\parallel}$ (**Fig.1a**) ($\mu_{\parallel}$=2.19± 0.06 kPa, $\mu_{\perp}$=2.44± 0.05 kPa, p=0.0107). In the GM, there was no significant difference between $\mu_{\parallel}$ and $\mu_{\perp}$ ($\mu_{\parallel}$= 2.08±0.07 kPa, $\mu_{\perp}$= 2.02± 0.07 kPa, p=0.1874). Also, the $\mu_{\perp}$ of WM was significant greater than (p<0.0001) the $\mu_{\perp}$ of GM, while no significant difference between the $\mu_{\parallel}$ of WM and the $\mu_{\parallel}$ of GM. The μFA of WM (μFA=0.32±0.01) was significantly greater than the GM (μFA=0.24±0.00, p=0.0011), and the DTI FA of WM (DTI FA=0.45±0.01) was significantly greater than the GM (DTI FA=0.11±0.00). The mean μFA (**Fig.2**) increased from 0.27 to 0.35 across the WM voxel groups with different DTI FA for which the mean DTI FA increased from 0.26 to 0.54 with a significant difference across all groups (Tukey-Kramer post-hoc analysis with 0.05 confidence level). There were significant differences between the μFA and DTI FA (p<0.0001) over all groups, and the DTI FA (**Fig.2**) was significantly greater than the μFA in all WM voxel groups except for the group with DTI FA values between 0.2 and 0.3, for which the μFA was less than the DTI FA. $\mu_{\perp}$ (**Table 1**) was significantly greater than $\mu_{\parallel}$ in all the WM structures (ANOVA, p=0.0015). The shear moduli of corpus callosum were significantly greater than the shear moduli of internal capsule (p=0.0339), while no significant difference between corona radiata and internal capsule or corpus callosum. In addition, the DTI FA and μFA in the corona radiata were significantly lower than in the internal capsule and corpus callosum (p=0.0161), which is consistent with rheological measurements of porcine brain [3].

Conclusions

This study of the anisotropic viscoelasticity of human brain tissues *in vivo* using a novel combination of MRE and DTI showed that white matter was slightly stiffer than grey matter. There was significantly higher anisotropy of shear moduli for WM compared to GM, and the $\mu_{\perp}$ of WM is significantly higher than that of GM. This is consistent with the anisotropic structure of WM consists of neural fibres as well as the structural anisotropy measured by the DTI FA. In particular, $\mu_{\perp}$ was significantly greater than $\mu_{\parallel}$ over all WM voxels as well as at the WM structures, which is consistent with the rheological measurements for the corpus callosum of porcine brains[3]. In addition, there was an increase of elasticity anisotropy in the WM voxels simultaneous to the increase of structural anisotropy measured by DTI FA. We quantified the shear moduli and their anisotropy in various WM structures, the results of which agree with direct rheological measurements [3]. The combination of MRE and DTI is a reliable and promising methodology for mapping the anisotropic viscoelasticity of human brains.

References

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