The influence of tension-compression switches on brain anisotropic modelling

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I. INTRODUCTION

Finite element (FE) models are vital for investigating the mechanics of traumatic brain injury (TBI). One important aspect of FE head models is brain material modelling. Despite not being universally agreed, a large body of brain tissue experiments reports the white matter (WM) to be mechanically anisotropic, consisting predominantly of aligned axonal fibers [1]. Employing the Gasser-Ogden-Holzapfel (GOH) continuum approach, these fibers act as reinforcements that require a tension-compression 'switch' to eliminate their mechanical contribution during compressive strain [2-4].

However, inconsistency exists in tension-compression switches regarding the parameters of these switches, and the treatment of fiber compression leaves their influence on WM responses unclear, while prior work has examined the switch from a theoretical perspective [4, 5]. The aim of this study is to investigate the influence of different tension-compression switch schemes.

II. METHODS

Finite element head model with embedded fiber orientation

The FE head model was developed by Kleiven (2007) [6]. To enable the anisotropic brain material modelling, the degree of anisotropy [characterized by the fractional anisotropy value (FA), Fig. 1A] and fiber orientation (Fig. 1B) delineated by diffusion tensor imaging (DTI) were coupled with the FE model on an element basis [3].

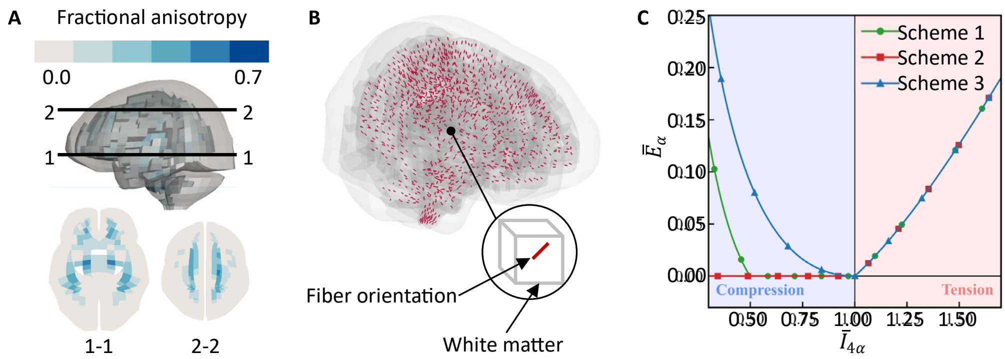

Fig. 1. (A) Axial cross-sections of the FE brain model reconstructed from DTI data; (B) Isometric view of the finite element model with embedded fiber orientation (red); (C) The plot of the function $\bar{E}_\alpha$ in relationship to $\bar{I}_{4\alpha}$ (κ = 0.2732). $\bar{I}_1 = \bar{I}_{4\alpha} + 2/\sqrt{\bar{I}_{4\alpha}}$, according to the incompressibility constraint.

Implementation of tension-compression switches in GOH model

Following GOH model [2], we adopt the decoupled form of a Helmholtz free-energy function to describe the WM:

$$\boldsymbol{\Psi} = K\left(\frac{J^2-1}{4} - \frac{1}{2}\ln J\right) + \frac{G}{2}(\bar{I}_1 - 3) + \frac{k_1}{2k_2}[\exp(k_2\bar{E}_\alpha^2) - 1]$$

where K represents the bulk modulus, J is equal to the determinate of the deformation gradient, G represents the shear modules, $\bar{I}_1 = tr\bar{\boldsymbol{C}}$ is the first invariant of the isochoric right Cauchy-Green strain tensor $\bar{\boldsymbol{C}}$. The quantity $\bar{E}_\alpha$ is used to calculate fiber contribution. $\bar{I}_{4\alpha}$ is equal to the square of the isochoric stretch in fiber orientation and can be used to calculate fiber strain $\bar{\varepsilon}_\alpha = \sqrt{\bar{I}_{4\alpha}} - 1 + (\sqrt{\bar{I}_{4\alpha}} - 1)^2/2$. k_1 and k_2 describe the fiber stiffness. Dimensionless parameter κ relates to FA values. Detailed descriptions can be found in pervious study [3].

This study focused on three tension-compression switch schemes: scheme 1 employed $\bar{E}_\alpha$ as the switching parameter while scheme 2 and 3 instead used $\bar{I}_{4\alpha}$. Scheme 2 treat $\bar{E}_\alpha$ as $\kappa(\bar{I}_1 - 3)$ when fiber was in compression

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($\bar{I}_{4\alpha}$- $\bar{E}_{\alpha}$ curves in Fig. 1C). These schemes were implemented in LS-DYNA via user-defined material.

Scheme 1: $$\bar{E}_{\alpha} = \begin{cases} \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)(\bar{I}_{4\alpha} - 1), & \bar{E}_{\alpha} > 0 \\ 0 \quad , & \bar{E}_{\alpha} \leq 0 \end{cases}$$

Scheme 2: $$\bar{E}_{\alpha} = \begin{cases} \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)(\bar{I}_{4\alpha} - 1), & \bar{I}_{4\alpha} > 1 \\ 0 \quad , & \bar{I}_{4\alpha} \leq 1 \end{cases}$$

Scheme 3: $$\bar{E}_{\alpha} = \begin{cases} \kappa(\bar{I}_1 - 3) + (1 - 3\kappa)(\bar{I}_{4\alpha} - 1), & \bar{I}_{4\alpha} > 1 \\ \kappa(\bar{I}_1 - 3) \quad , & \bar{I}_{4\alpha} \leq 1 \end{cases}$$

Boundary conditions

All models were exposed to an impact loading measured by instrument mouthguards from collegiate football players known to induce concussion [7].

III. INITIAL FINDINGS

The tension-compression switch schemes affected WM deformation. For one random selected element, scheme 2 exhibited the highest maximum 1st principal Green-Lagrange strain (MPS), while scheme 3 yielded the lowest (Fig. 2A), resulting in a maximum inter-scheme MPS difference of 33.8%. Interestingly, the MPS yielded by scheme 1 was slightly higher than that of scheme 3, with a difference of 12.3%. Setting scheme 2 as the baseline, minimum fiber strain value in scheme 1 and 3 lowered by 28.7% and 36.5%, respectively (Fig. 2B).

At a representative time point (i.e., 27 ms) for the illustrative impact, when using $\bar{E}_{\alpha}$ as the switch parameter (i.e., scheme 1), fibers in 218 WM elements were in compression. When using $\bar{I}_{4\alpha}$ as the switch parameter, the number of WM elements with compressed fibers were 189 for scheme 2 and 183 for scheme 3 (Fig. 2C).

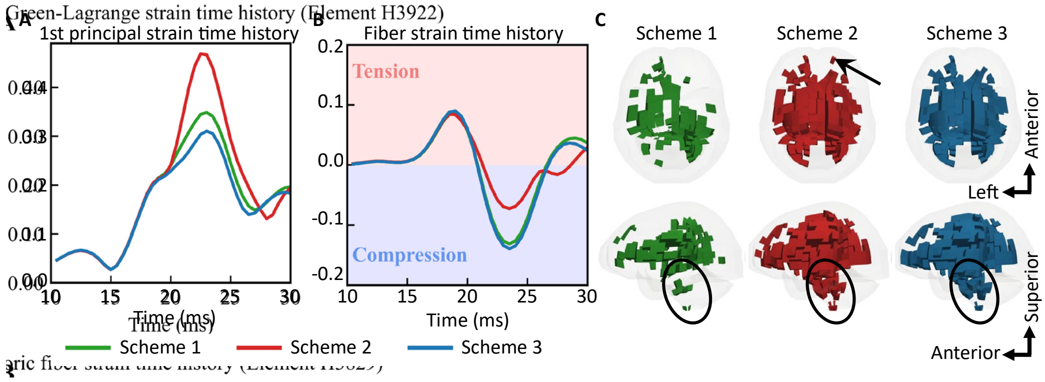

Fig. 2. (A) Time history curves at the first peak (i.e., 10-30 ms) of the first principal strain; (B) Time history curves at the first peak (i.e., 10-30 ms) of the fiber strain; (C) Superior and lateral view of WM with the embedded fiber in compression.

IV. DISCUSSION

The current study presented a theoretical elaboration and numerical implementation of three candidate tension-compression switches in the GOH model to model the mechanical anisotropy of WM in impact simulations. Our results verified that both the tension-compression switch parameter (($\bar{E}_{\alpha}$ vs. $\bar{I}_{4\alpha}$) and the treatment of fiber's mechanical contribution in compression (0 vs. $\kappa(\bar{I}_1 - 3)$) affected the brain responses. Existing anisotropic FE head models are developed in different commercial computational software, e.g., scheme 1 in Abaqus and scheme 2 in LS-Dyna. Caution is warranted when comparing injury risk responses based on fiber strain derived from software that employs different switching criteria.

V. ACKNOWLEDGEMENTS

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VI. REFERENCES

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