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The influence of anisotropy on brain injury prediction

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ABSTRACT

Traumatic Brain Injury (TBI) occurs when a mechanical insult produces damage to the brain and disrupts its normal function. Numerical head models are often used as tools to analyze TBIs and to measure injury based on mechanical parameters. However, the reliability of such models depends on the incorporation of an appropriate level of structural detail and accurate representation of the material behavior. Since recent studies have shown that several brain regions are characterized by a marked anisotropy, constitutive equations should account for the orientation-dependence within the brain. Nevertheless, in most of the current models brain tissue is considered as completely isotropic. To study the influence of the anisotropy on the mechanical response of the brain, a head model that incorporates the orientation of neural fibers is used and compared with a fully isotropic model. A simulation of a concussive impact based on a sport accident illustrates that significantly lowered strains in the axonal direction as well as increased maximum principal strains are detected for anisotropic regions of the brain. Thus, the orientation-dependence strongly affects the response of the brain tissue. When anisotropy of the whole brain is taken into account, deformation spreads out and white matter is particularly affected. The introduction of local axonal orientations and fiber distribution into the material model is crucial to reliably address the strains occurring during an impact and should be considered in numerical head models for potentially more accurate predictions of brain injury.
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1. Introduction

Traumatic Brain Injury (TBI) can occur when the head suddenly and violently hits an object and the resulting internal forces produce damage to the brain, disrupting its normal function. A person suffering from TBI experiences devastating effects on overall neurological function (Langlois et al., 2003). Moreover diffuse axonal injury (DAI), one of the most severe forms of TBI, is a common cause of death or permanent disability worldwide (Faul et al., 2010). Improved diagnostic tools and protective measures can help to reduce the incidence and seriousness of injuries; a better understanding of TBI is therefore needed. Investigation of the relation between mechanical load and the resulting injury can lead to the development of brain injury criteria. To this effect, finite element (FE) modeling is a powerful technique through which impact-tolerance can be studied. Mechanical variables such as pressure, maximal principal strain or invariants of the strain and stress tensor can be extracted and they can be used as injury prediction measures (Bain and Meaney, 2000; Morrison et al., 2006; La Placa and Prado, 2010).

However, in order to correctly use tissue-level brain injury criteria to interpret head model simulation results, the tissue deformations need to be predicted accurately and the used models must provide an appropriate level of structural detail. Contemporary FE models should contain both viscoelastic material descriptions to account for the time-dependent behavior of brain tissue as well as anisotropic characteristics to capture the orientation-dependency of the brain.

Previous experimental studies on porcine brains have shown that the stiffness of the tissue is directionally dependent, especially at large strains. Arbogast and Margulies (1998) tested tissue samples of the brainstem for various shear strain directions and found differences in the shear modulus of up to 20%. In a study reported by Prange and Margulies (2002), the response of the brain was evaluated in the superior–inferior and transverse direction. A variation of stiffness was found for different regions of cerebral white matter and differences in the shear modulus were 30% and 92% for the corona radiata and the corpus callosum respectively. Also, Hrapko et al. (2008) studied the shear response of the corona radiata and observed that the maximum shear modulus was up to 54% higher than the minimum shear modulus of the same tissue sample depending on the strain magnitude. Ning et al. (2006) conducted shearing tests in different directions on the brainstem and fitted a fiber-reinforced constitutive model with a quadratic reinforcing strain energy function to accommodate the

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anisotropic behavior of the brainstem. In a study performed by Colgan et al. (2010), anisotropy was incorporated in an FE head model and it was shown that tissue stress and strain states are significantly affected. However, these observations were based on extremely high magnitude loading conditions (i.e., angular acceleration of 175 krad/s^2) and a large difference in stiffness between the brainstem and the remaining brain tissue.

Since a correlation between axonal structural anisotropy and mechanical response is undeniable, any additional information of micro-scale axonal distribution may be useful to enhance axonal strain calculations to better predict brain injuries. Directional measurements of water diffusion in soft tissue using diffusion tensor imaging (DTI) can provide information about tissue architecture. This important information can then be coupled with a FE head model, leading to the integration of heterogeneous and anisotropic properties into the brain material model (Giordano and Kleiven, 2013). White matter structure will likely influence the strain measures. However, for the time being, only a few studies have evaluated its microstructural distribution into injury analysis (Colgan et al., 2010; Chatelin et al., 2011; Wright and Ramesh, 2012). For example, Colgan et al. (2010) modeled the brain as a transversely isotropic hyperelastic material and studied the effects of applying high rotational acceleration/deceleration into the shear strain response. Chatelin et al. (2011) instead proposed a post-processing approach to couple FE tissue-level stresses from an isotropic brain model with structural anisotropy from DTIs and reconstructed axonal elongation for two head trauma cases presenting different injury levels. Wright and Ramesh (2012) incorporated the structural orientation of the neural axons into a 2D FE model, finding that the degree of injury predicted by the model was direction-dependent. However, none of these studies considered a 3D anisotropic FE model with distributed axonal fiber orientation.

In the current work, a previously developed head model (Kleiven, 2007) is improved to investigate the influence of fully anisotropic brain tissue on the mechanical internal head response. Brain matter is assigned a hyper-viscoelastic anisotropic behavior: axonal fiber pathways are detected with DT imaging and the degree of anisotropy of axonal bundles and their orientation are integrated into the constitutive law (Giordano and Kleiven, 2013). Through the use of LS−DYNA®, the kinematics of a concussive impact between two football players is finally applied both to the isotropic and the new anisotropic model and a comparison between the mechanical responses of brain tissue is performed. The aim of the work is to investigate the influence of anisotropy on the mechanical behavior of the brain.

2. Material and methods

2.1. Head model

The finite element model used in this study was developed at the Royal Institute of Technology (KTH) in Stockholm by Kleiven (2002) using the LS−DYNA® software. The original version is a detailed 3D model of the head developed for looking into the effects of impact loads on the head. It has shown good correlation with experiments of skull-brain relative motion (Kleiven and Hardy, 2002; Kleiven, 2006b; Giordano and Kleiven, 2013), intra-cerebral acceleration (Kleiven, 2006a), skull fracture (Kleiven, 2006a) and intra-cranial pressure (Kleiven and Von Holst, 2002; Kleiven, 2006b). As shown in Fig. 1, the KTH head model includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid (CFS), eleven pairs of the largest parasagittal bridging veins and a simple neck with the extension of the spinal cord and the dura mater. The differentiation between gray and white matter and the inclusion of the ventricles are also implemented. All parametrical choices concerning the geometry of the head are based on a detailed analysis of previous studies (Kleiven, 2007).

In the present work, the original model is extended with the purpose to account for the anisotropy of brain tissue. The same constitutive laws as in the original KTH model are used for the various anatomical components of the head (Table 1) with the exception of the brain where the tissue is assigned a hyper-viscoelastic fiber-reinforced anisotropic model, according to the formulation proposed by Gasser, Ogden and Holzapfel (GOH) for modeling arterial layers with distributed collagen fiber orientations (Gasser et al., 2006). The new material model

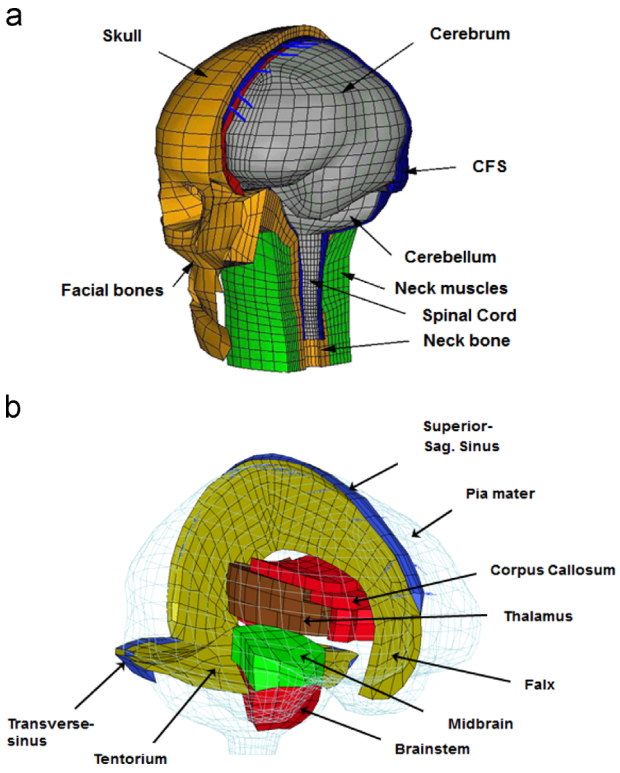

Fig. 1. Finite element model of the human head. Top: isometric view of the head model with brain exposed. Bottom: internal view of the head model.

Table 1
A summary of the properties of the head model components used in this study. The capital letter *K* represents the bulk modulus while EA means force/unit strain.

Tissue	Young's modulus [MPa]	Density [kg/dm^3]	Poisson's ratio
Outer compact bone	15 000	2.00	0.22
Inner compact bone	15 000	2.00	0.22
Porous bone	1000	1.30	0.24
Neck bone	1000	1.30	0.24
Brain tissue (Model 1)	GOH (k=0.3333)		
Brain tissue (Model 2)	GOH (k depending on FA)		
Cerebrospinal fluid	K=2.1 GPa	1.00	–
Sinuses	K=2.1 GPa	1.00	–
Dura mater	31.5	1.13	0.45
Falx	31.5	1.13	0.45
Tentorium	31.5	1.13	0.45
Pia mater	11.5	1.13	0.45
Bridging veins	EA=1.9 N	–	–

is indeed capable of integrating information on tissue composition accounting for the internal load carrying mechanisms of the individual constituents. Moreover, the dispersion of the fibers is taken into account thanks to a scalar structure parameter, k, that enters the hyperelastic formulation. The hyperelastic strain energy potential of brain tissue is therefore defined as

$$W=\frac{G}{2}(\tilde{I}_1-3)+K\left(\frac{J^2-1}{4}-\frac{1}{2}\ln(J)\right)+\frac{k_1}{2k_2}(e^{k_2\langle\tilde{E}_\alpha\rangle^2}-1) \quad (1)$$

where the third term on the right-hand side was based on the GOH form (Gasser et al., 2006) with only one fiber family

$$\tilde{E}_\alpha=k(\tilde{I}_1-3)+(1-3k)(\tilde{I}_{4\alpha}-1) \quad (2)$$

W represents the strain energy per unit of reference volume, G is the shear modulus, $\tilde{I}_1$ denotes the first invariant of the isochoric part of the Cauchy–Green strain tensor, K defines the bulk modulus, J is equal to the determinant of the deformation gradient and represents the volume ratio, k_1 and k_2 are the parameters related to the fiber stiffness.

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