

## A Comparison between Simulated and Measured Human Brain Response under Mild Acceleration.

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**Abstract** Due to the limitation of in vivo studies on human brain, finite element modelling is an important tool to study brain injury. Although head finite element models are developed to mimic brain responses to impact conditions, there is a need to consider how closely these models simulate actual brain. In this work, the effect of brain material models, geometry of the head and the brain-skull connection is investigated for a mild deceleration of the head by comparing the strain response of brain models with the strains measured using magnetic resonance imaging. The comparison considered the SIMon, THUMS, and GHBMC finite element head models. The simulated strain responses from all the models agreed with the experimental peak strain locations, but were lower in magnitude. While the brain model and brain-skull connection of the SIMon model was different from the THUMS and the GHBMC models, the strain response of all the models was similar. For the simulated impacts considered here (~40 ms duration), the neck contributed measurably to the rotational acceleration of the head and, in turn, brain strain.

**Keywords** brain deformation, finite element head models, head kinematics, validation

### I. INTRODUCTION

Brain injury is a major health concern receiving increased attention, particularly in sport [1]. The brain can be injured in recreational activities, car accidents or falls. Understanding the mechanism of such injuries is the first step for preventing them. To study brain injury, insight into the mechanical response of the brain to traumatic loadings is needed. Given the difficulty in studying human brain in vivo, current tools to study brain injury often involve post-mortem human subject (PMHS) and animal testing [2-4], anthropomorphic test devices (ATDs) [5,6], and computational models [7,8].

Due to the need to study the brain responses to impact loads, the finite element method is receiving more attention as a tool to study head trauma. There has been a continued effort to develop and improve the finite element biofidelic head models over the last decade, including the Total Human Model for Safety [THUMS], Global Human Body Models Consortium [GHBMC], Simulated Injury Monitor [SIMon], Kungliga Tekniska Högskolan [KTH] head model, The Dartmouth Head Injury Model [DHIM], and An Anatomically Accurate Finite Element Brain Model ABM [9]. The use of material models for the constitutive parts, geometric representations, and contact definitions are important features of each model that determine how they simulate the human head. Validation of finite element [FE] models is an important step in model development. The validation process of the head models usually consists of measuring the accuracy of the kinematic responses of the model, brain pressure, and displacement of the brain with respect to the skull using PMHS tests. PMHS data in different impact directions are used to validate the head acceleration at the centre of gravity, impact force and duration of the simulated impact with FE models [10,11]. Intra cranial pressure experiments [12] and local brain motion measurements with neutral density targets (NDTs) [2] are used to validate the brain responses of finite element models. These experimental tests bring insight into the brain response from impact, but these validation methods have limitations. Correct pressure response of the brain does not ensure correct deformation [13]. Direct measurement of brain strain is not possible using NDTs [14], and since these motion measurements are localised, the global pattern of brain movement is not discernable. Furthermore, the mechanical response of human brain may change significantly after death [15].

Although in vivo strain measurements of the human brain [15-17] have been performed under mild acceleration and the FE head models are developed to predict the brain behaviour in impact conditions with high acceleration, they can be effective in validating numerical head models. Recently, the DHIM model was

validated by comparing the area fraction of radial circumferential strain ( $\epsilon_{rf}$ ) exceeding three threshold levels, measured in live human brain under rotational acceleration [16,18]. There has been no work, however, comparing strain distribution from the numeric simulations with these experimental results.

The objective of this study was to compare the brain deformation of the three widely used FE head models (GHBMC, THUMS, and SIMon) with the deformation of live human brain under mild acceleration. The effects of neck and brain material properties on brain deformation are also investigated in this work.

## II. METHODS

To compare the simulated and human brain deformation, the human experiment was recreated in LS DYNA. In the experimental setup described in [15], the volunteer's head (3 subjects, healthy adults 22–39 years old, 2 males, 1 female) was placed inside a rigid frame that constrained the motion of the head to pivot in the direction of neck flexion-extension. The frame was released to fall 2 cm on to a rigid stop, while the head was caught by a suspension system to decelerate the head with a peak deceleration of 20 to 30 m/s<sup>2</sup> over about 40 ms. This motion was repeated 144 times to obtain one sagittal plane and one axial plane magnetic resonance image [MRI]. The linear acceleration of each participant was measured after scanning, where only the x-linear component (Fig. 1) was reported (rotational acceleration was not measured). Vertical and horizontal tag lines were acquired to visualize the motion of the brain. The intersection of these lines created 192 spatial points on the brain. These spatial points were used to define the vertices of a triangular mesh in a reference image and deformed image. The local deformation gradient tensor  $F$  was estimated by comparing the deformed and reference meshes, from which the Lagrangian strain tensor was obtained.

A 50<sup>th</sup> percentile male THUMS model version 4.02 [19], a 50<sup>th</sup> percentile male GHBMC model student version 4-5 [9], and a SIMon head model<sup>1</sup> [20] were used in this study. All the simulations were performed using LS DYNA (version mmp s R8.1.1). Since only one component of acceleration was measured in the experiment, the measured acceleration could not be used to reconstruct the impact for the simulations. The experiment was simulated, therefore, by releasing the head from 2, 3, and 4 cm onto a soft surface (Fig 1). The impact surface was modelled as a 500×500×50 mm foam pad using a FU\_Chang foam material model (Mat 83)[21]. The nodes of the bottom surface of the foam were fixed in all 3 degrees of freedom. To model the contact between the head and the foam, automatic surface to surface contact with a coefficient of friction 0.1 was used. Gravity was applied to the head while the rest of the body was unconstrained. Since the experiment did not report the compliance of the suspension system decelerating the head, the simulation drop height onto the foam surface was adjusted until the simulated head deceleration matched that reported in the experiment. A 4 cm head drop in the simulation provided the best agreement with the experiment (2.5 g). Since the SIMon head model does not include a neck or body, the accelerations from the THUMS model were applied to the SIMon skull to simulate the head drop.

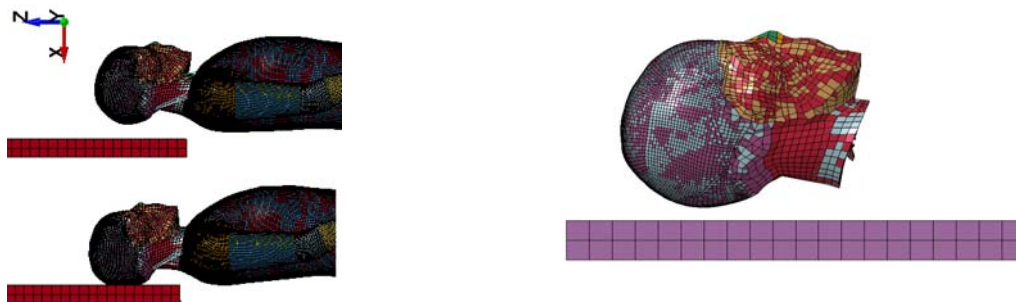


Fig. 1. THUMS Full body and isolated head simulation of the head drop on a foam surface.

The stress-strain response of the brain tissue is time dependent. The THUMS, GHBMC, and SIMon all use viscoelastic models for the brain material to address this time dependency. To compare how the brain deformation is affected by the brain material models, the response of the THUMS model was compared using the SIMon and GHBMC brain material models (see Table I).

In many head impact simulations and ATD tests [22–24] only the head or head and neck are present, while the contribution of rest of the body is neglected. Some have argued that the impact duration is sufficiently short

<sup>1</sup> This model was created on Feb 2008.

that the rest of the body does not affect the kinematics of the head. To compare the effect of the body connection to the head, the THUMS model was dropped from 2, 3, and 4 cm with and without the body connected.

TABLE I  
VISCOELASTIC MATERIAL PROPERTIES USED FOR THE THUMS, SIMon, AND GHBMC MODELS ( $\rho$  IS DENSITY,  $K$  IS BULK MODULUS  $G_0$  AND  $G_1$  ARE INSTANTANEOUS AND LONGTERM SHEAR MODULUS, AND  $\beta$  IS THE DECAY CONSTANT.

| Material Model                              | Parameters                                                                                                                                                                                                                                                                                                                                           |
|---------------------------------------------|------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| <b>THUMS brain model</b><br>(Maxwell model) | $\rho = 1 \text{ g/mm}^3$ $K = 2160 \text{ MPa}$ $G_1 = 1200 \text{ Pa}$ , $\beta = 80$ , $G_0 = 6000 \text{ Pa}$                                                                                                                                                                                                                                    |
| <b>SIMon brain model</b><br>(Maxwell model) | $\rho = 1.04 \text{ g/mm}^3$ $K = 558.47 \text{ MPa}$ $G_1 = 928 \text{ Pa}$ , $\beta = 16.95$ , $G_0 = 1662 \text{ Pa}$                                                                                                                                                                                                                             |
| <b>GHBMC brain model</b><br>(Kelvin model)  | $\rho = 1.06 \text{ g/mm}^3$ $K = 2190 \text{ MPa}$<br>$G_1 = 7500 \text{ Pa}$ , $\beta = 0.0125$ , $G_0 = 1500 \text{ Pa}$ (Cerebrum white matter)<br>$G_1 = 1200 \text{ Pa}$ , $\beta = 0.0125$ , $G_0 = 6000 \text{ Pa}$ (Cerebrum gray matter, cerebellum)<br>$G_1 = 2400 \text{ Pa}$ , $\beta = 0.0125$ , $G_0 = 12000 \text{ Pa}$ (Brain stem) |

### III. RESULTS

Table II compares the THUMS and GHBMC full body models dropped from 2, 3, and 4 cm as depicted in Fig. 1. Fig. 2 shows the linear acceleration,  $a_x(t)$ , rotational acceleration,  $\alpha_y(t)$ , and the head-foam contact force,  $F(t)$ , for the THUMS and GHBMC 4 cm drop impacts as a function of time (head is released at  $t=0$ ).

A representative experimental linear acceleration is included in Fig. 2 for comparison [15]. Both FE models show good agreement with experiment. Rotational acceleration from the simulations is also presented in Fig. 2. Note the differences in rotational acceleration between the THUMS and GHBMC models. This is due to a difference in neck posture and stiffness of the two models which explains the difference between their contact forces.

TABLE II  
PEEK LINEAR DECELERATION, DECELERATION DURATION, AND CONTACT FORCES FOR THE THUMS, AND GHBMC MODELS IN 20, 30, and 40 MM DROP SIMULATIONS.

|       | Drop height<br>(cm)                             | 2   | 3   | 4   |
|-------|-------------------------------------------------|-----|-----|-----|
| THUMS | Peak linear deceleration<br>(m/s <sup>2</sup> ) | 22  | 24  | 26  |
|       | Deceleration duration<br>(ms)                   | 40  | 40  | 42  |
|       | Maximum contact force<br>(N)                    | 150 | 174 | 199 |
|       |                                                 |     |     |     |
| GHBMC | Peak linear deceleration<br>(m/s <sup>2</sup> ) | 19  | 22  | 23  |
|       | Deceleration duration<br>(ms)                   | 41  | 40  | 34  |
|       | Maximum contact force<br>(N)                    | 120 | 130 | 135 |
|       |                                                 |     |     |     |

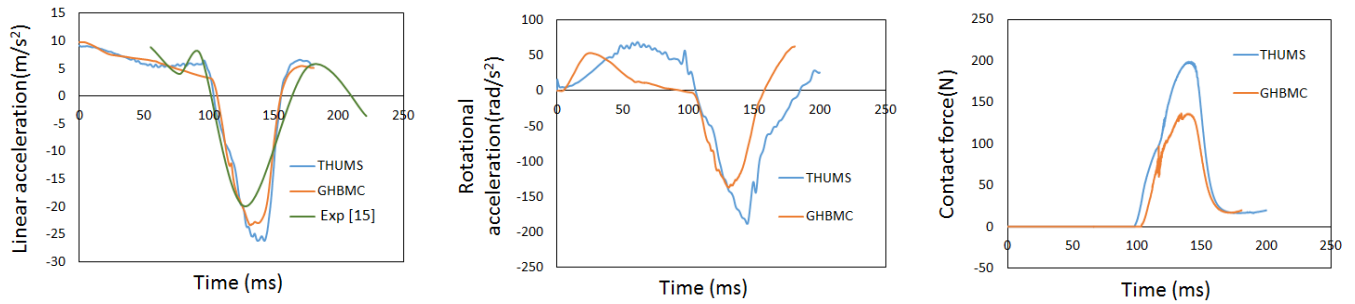


Fig. 2. THUMS and GHBMCM full body simulation results for 4 cm head drop.

The experimental brain strain response is compared with the FE models in Tables III and IV in the sagittal and axial planes, respectively. While the impact conditions are identical, the strain magnitudes from the models are not, and are uniformly lower than experiment. Accordingly, the colour scale is tailored for each model to allow comparison. The maximum principal strain (MPS) time history at three brain locations (Fig. 3) are compared for all the models in Fig. 4. Location 1, at the top of the brain, is the location of peak MPS for the GHBMCM model. Location 2 is at the centre of contact between the head and foam surface. Location 3 is on the boundary of the cerebrum and cerebellum, where the peak MPS happens for the THUMS model.

Before the impact, at  $t=74$  ms, gravity is the only external force acting on the head. For all the models, the strain magnitude before the impact is much smaller than the MRI results. Recall the volunteer's head was placed in a frame which constrains head movement. The volunteer is also aware of the impact. This anticipation can affect neck muscle activation and influence rotational head movement before impact. Since strain is sensitive to rotational acceleration [7,25,26], which was not measured in the MRI experiments, differences in strain between the models and MRI experiments could be due to rotational effects. Similarly, differences in rotational acceleration of the THUMS and GHBMCM models before the impact (Fig. 2) may explain the different strain patterns in these models. The GHBMCM model, for instance, shows a pre-impact peak strain at  $t=40$  ms (Fig.4). The different strain distributions of the THUMS and SIMon models are not due to head kinematics since they were given the same acceleration. The difference between the THUMS and SIMon models brain response is likely due to differences in the skull-brain connection and brain properties. Dura, arachnoid trabeculae, pia matter and cerebral spinal fluid (CSF) comprise the layers between the skull and the brain, and are responsible for the movement of the brain relative to the skull. While these layers are modelled separately in the THUMS and GHBMCM models with different material models and parameters, they are combined and defined as a single linear viscoelastic material in the SIMon model.

During the impact, all the models reach their peak strain near  $t=150$  ms (Fig. 4). While lower in magnitude, the simulated response shows good agreement with experiment for peak strain locations (Tables III and IV). Tension near the site of impact and compression in the opposite site of the impact in the X direction is captured in the MRI. FE models show this result in the axial plane but not in the sagittal plane. The reason is that while the maximum tension and compression strain is close to centre of the brain in the MRI, the simulated peak strain is closer to the boundaries, and the location of the sagittal plane is close to the centre of the brain [15]. Differences in the location of  $\max \epsilon_x$  between the models may be due to the different geometry of the head that can change the location of contact between the brain and the skull. In the Z direction, the MRI shows the maximum compression strains at the top left and bottom right side of the cerebrum and the maximum tension strain at the top right and bottom left side of the cerebrum. These observations are also true for the THUMS and GHBMCM models while for the SIMon model, the maximum compression strains were shifted towards the center of the brain. The magnitude of strain in the Y direction ( $\epsilon_y$ ) is insignificant compared to the other two directions. MRI and the THUMS model show small compression and tension in the front and rear part of the brain respectively (light blue and yellow areas). Given the small magnitude, however,  $\epsilon_y$  does not contribute to validating the FEM. The FE models show more consistent strain distribution with the MRI in shear. Negative  $\epsilon_{xz}$  at the top and bottom of the brain, positive  $\epsilon_{xz}$  in the middle section of the brain, and the negative and positive  $\epsilon_{xy}$  at the right and left side of the brain respectively are observed in both MRI and FE models.

The THUMS model had higher rotational acceleration than the GHBMC model which led to higher strain magnitudes. An important difference between the simulations and experiment is that while MRI shows the cerebellum and brain stem have high strain, low strains were observed in this area in the simulations. The simulated brain stem strains appear to be transferred to the boundary of cerebrum, and cerebellum. This may be due to the relatively simple isotropic material models used in the simulations that do not describe anisotropic behaviour of the brain in these areas. It has been suggested, for instance, that the brain stem and the corpus collegium have highly anisotropically oriented axons [27].

After the impact, at  $t=175$  ms, the GHBMC and the SIMon models are strain free. While the THUMS model shows strain at  $t=175$  ms, it is still small compared to experiment. The difference in time dependence between the THUMS and SIMon models is likely related to their decay constants, which is larger for the THUMS model (see Table I). The difference between the THUMS and GHBMC models may be due to the different rotational acceleration of the head, as will be discussed below.

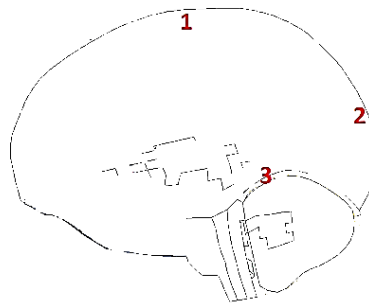


Fig. 3. Selected locations of the brain on the MPS over time plots.

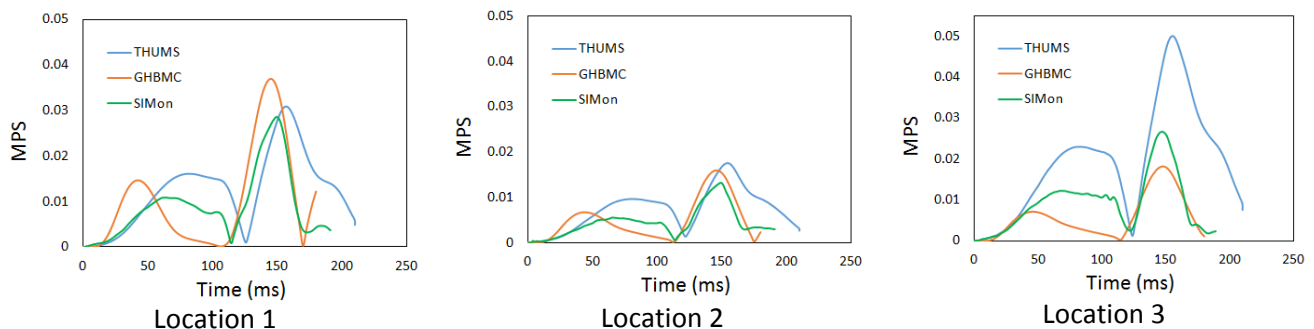


Fig. 4. Maximum principal strain vs. time for THUMS, GHBMC, and SIMon.

### Brain material model

The foregoing has compared different FE models with their unique geometries and material properties. To consider the effect of the brain material properties, independent of model geometry, Fig. 5 compares the MPS using the brain material properties of the THUMS, GHBMC and SIMon models (Table I) with the geometry of the THUMS model. (All geometry and remaining material properties in the THUMS model were not changed.) Fig. 5 shows that the response of the GHBMC and THUMS brain material models were similar, while the SIMon brain material model has higher strain. Over the short duration and low strain rate ( $\dot{\epsilon}_1 < 10 \text{ s}^{-1}$ ) of these 4 cm drop impacts, the stiffness of the viscoelastic material is mostly governed by the instantaneous shear modulus, which is the same for the THUMS and GHBMC cerebrum material and lower for the SIMon brain material (Table I). Apparently, the single material used by the SIMon model to describe the multiple layers between the brain and skull provide an interface that is too stiff. A softer brain material model helps to accommodate a stiffer interface and provide a more representative response for the head [20].

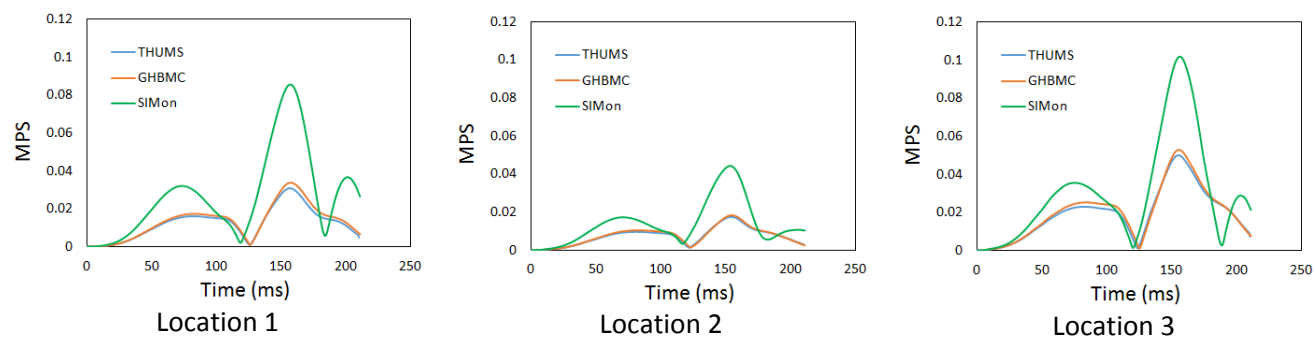


Fig. 5. MPS as a function of time for the THUMS model using the THUMS, GHBMC and SIMon brain material models.

TABLE III  
Table. 3. QUALITATIVE COMPARISON OF THE BRAIN STRAIN IN THE SAGGITAL PLANE FROM A 4 CM HEAD DROP ONTO A FOAM SURFACE USING THE THUMS, GHBMC, AND SIMON FE MODELS WITH EXPERIMENTAL DATA FROM MRI [15].

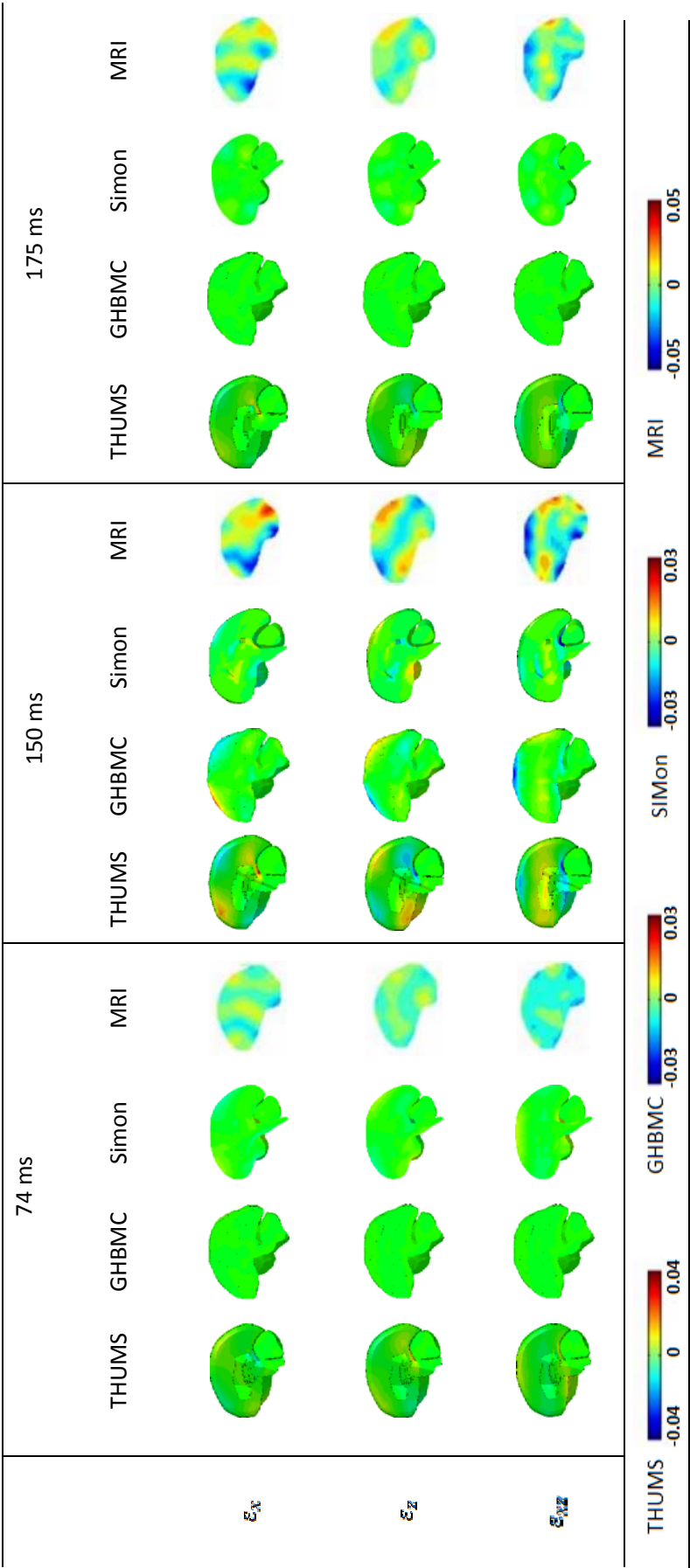
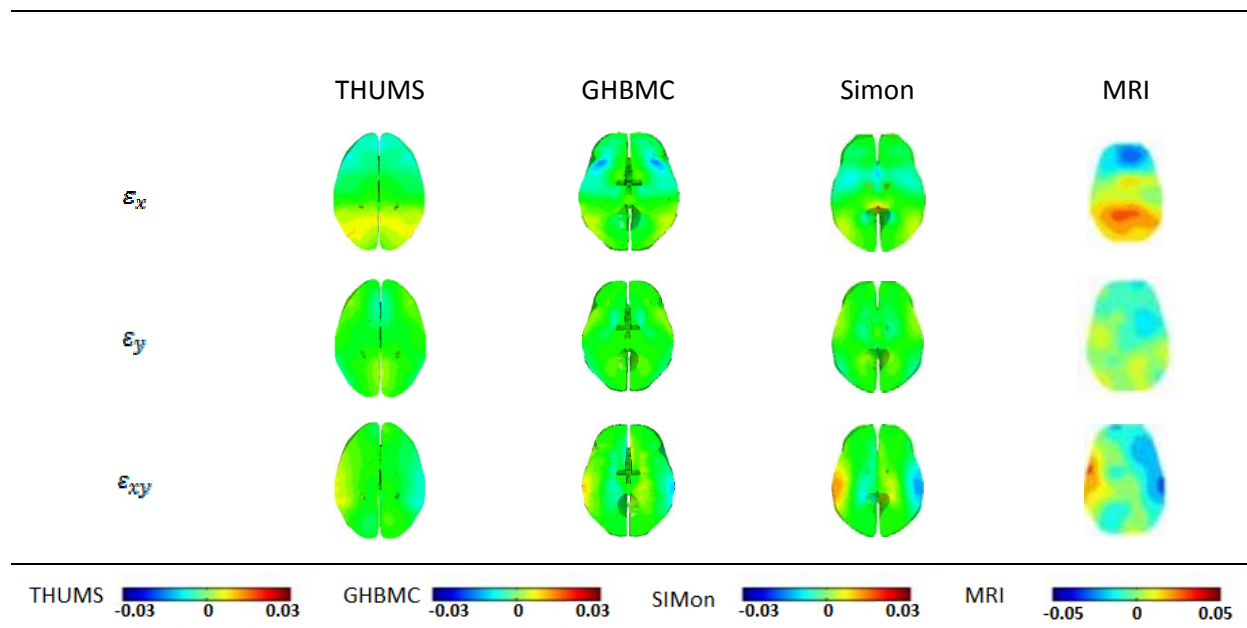


TABLE IV  
BRAIN STRAIN IN AXIAL PLANE FROM THUMS, GHBMCM, SIMon, AND MRI AT 150 MS.



### Isolated head simulation

To consider the effect of the neck, the THUMS head was dropped from 2, 3, and 4 cm without connection to the body. Peak linear and rotational accelerations as well as contact forces are presented in Table V (compare with Table II). Fig. 6 shows the linear acceleration,  $a_x(t)$ , rotational acceleration,  $\alpha_y(t)$ , and the head-foam contact force,  $F(t)$ , for the 4 cm drop impacts as a function of time (head is released at  $t=0$ ) for the THUMS head with and without connection to the body. Since, in the head only simulation, there is no constraint on the movement of the head due to the attachment to the body, the rotational acceleration changes significantly while the linear acceleration stays constant.

TABLE V  
PEAK LINEAR DECELERATION, DECELERATION DURATION AND CONTACT FORCES FOR THUMS HEAD ONLY SIMULATION.

| Drop height (mm)                             | 20  | 30  | 40  |
|----------------------------------------------|-----|-----|-----|
| Peak linear deceleration (m/s <sup>2</sup> ) | 24  | 28  | 31  |
| Deceleration duration (ms)                   | 45  | 41  | 39  |
| Contact force (N)                            | 200 | 228 | 250 |

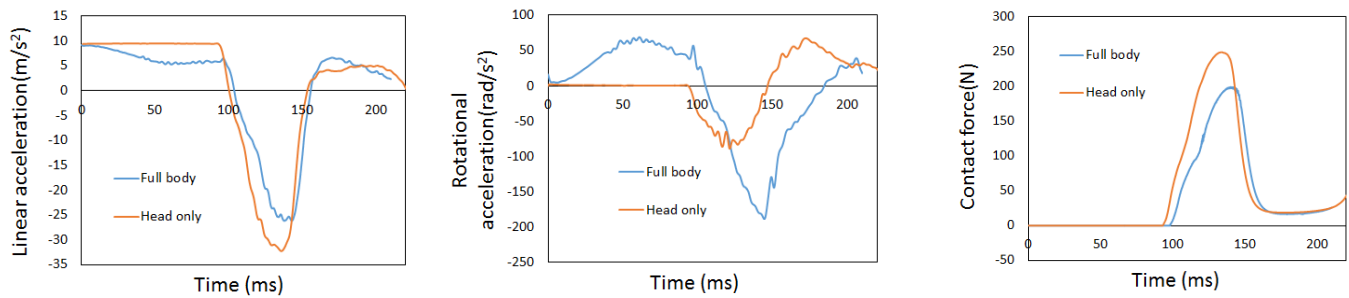


Fig. 6. THUMS head only and full body simulation dynamic results.

Fig. 7. compares the MPS for the full body and head only simulations of the THUMS model over time. The MPS before impact for the head only simulation was negligible compared to the full body simulation. The higher strain observed in the full body impact is likely due to its higher rotational acceleration (Fig. 6). After impact ( $t=160$  ms), the head only rotation is not damped by the inertia of the body. This may explain why the MPS of the head only increased while the full body simulation decreased.

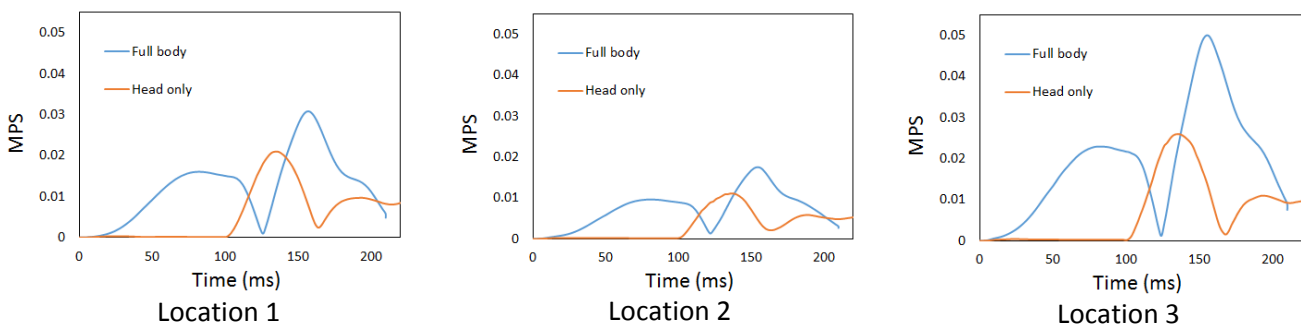


Fig. 7. MPS over time for full body and head only simulation

## I. DISCUSSION

The biofidelity of three head models (THUMS, GHBMC, and SIMon) were evaluated in this study using the brain strain measured with MRI. The differences between simulated results and the MRI can be divided in two categories; errors involved with the experimental testing of the volunteer's head, and the inability of the head models to replicate the human head. Since all three models were used to reconstruct the same impact, comparing the strains of these models with each other shows the differences that are due to the models.

Geometric representation of the constitutive parts, material models used for these parts, and contact definitions are features that control a model's biofidelity. To study the contribution of each of these parameters on the brain they should be varied independently. For example, the material model test in this work showed that the differences in strain of the brain in THUMS and GHBMC is not due to the different brain material models, while the differences in the THUMS and SIMon brain models is due to differences in the brain models. The material model test also showed that the SIMon brain is softer than the THUMS brain. Since the SIMon and THUMS models have the same kinematics, the lower strains in the SIMon model was due to the lower load transferred from the skull to the brain. Note that the impact reconstruction using the SIMon model was different from the other two models as the kinematics of the head was the input to the rigid skull versus dropping the THUMS and GHBMC heads on the soft surface. Whether this difference in the impact affects the brain deformation has not been investigated in this study. The THUMS and GHBMC models on the other hand had different kinematics in this impact, therefore the comparison of the brain-skull connection can't be concluded independently from their different brain responses.

To determine the effect of the kinematics of the head on the brain deformation, the THUMS head was separated from the rest of the body and used to model the same impact. The results showed the dominance of the rotational acceleration on controlling the brain strains compared to the linear acceleration. This result shows the need for in vivo studies involving pure rotational acceleration [17] to further validate the simulated brain deformation.

## II. CONCLUSIONS

The foregoing has compared three finite element models (THUMS, GHBM, and SIMon) with experimental in vivo brain strains from a low magnitude head acceleration. Although the finite element head models are designed to predict the behaviour of the brain in traumatic impacts, i.e., high acceleration/short duration, the THUMS, GHBM, and the SIMon models were able to show similar stress distribution compared to the MRI results. However, the magnitudes of peak strains were different between the models and MRI which may be due to differences in the rotational acceleration of the head, brain material model, and the skull-brain connection model. The constraints attaching the body to the head increase the rotational acceleration of the head which leads to more deformation of the brain.

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