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Effects of Head Morphology on Brain Strains Due to Impacts: A Numerical Approach

K. Gupta[1,2], P. Pavan[1] and U. Galvanetto[1,2]

[1]Department of Industrial Engineering, University of Padova
Padova, Italy
[2]Center of Studies and Activities for Space (CISAS) "G. Colombo", University of Padova
Padova, Italy

Abstract

Traumatic Brain Injury is a major public health concern, and recent research has highlighted the significant influence of head morphological variations on injury outcomes, under the same kinematics. Finite element head models are important tools for understanding injury biomechanics, but existing models often lack anatomical detail. This study presents the development of detailed anatomical male and female finite element head models and assesses tissue-level responses (maximum principal strain, and shear strain) under pure translational, rotational, and combined translational-rotational kinematics based on experimental data. Results demonstrate significant differences in peak strain values (max up to 46%) and varying peak strain locations between the two models, highlighting the impact of morphological variations on brain response. These preliminary results suggest a possible relationship between intracranial volume and strain values, emphasizing the importance of considering individual anatomical differences in traumatic brain injury assessment and prevention strategies.

Keywords: finite element analysis, computational mechanics, biomechanics, brain morphology, traumatic brain injury, impact analysis.

1 Introduction

Traumatic Brain Injury (TBI) is a significant contributor to injury-related mortality and disability, having substantial consequences on individuals and their families [1]. Posing a considerable economic burden on society and a threat to global public health,

TBI affected millions worldwide in 2019 [2], with an estimated 27.16 million new cases, 48.99 million prevalent cases, and 7.08 million years lived with disability (YLDs). The European Union alone faces approximately 7.7 million prevalent cases, 1.5 million hospital admissions, and 57,000 deaths annually due to TBI [3]. Affecting individuals across all age groups, including children, adolescents, and the elderly, TBI stems from various causes such as traffic accidents, falls, and sports-related injuries. Extensive research is underway to assess the risk of head injury and analyse the biomechanical factors involved in head impacts, utilizing various biomechanical tools and methodologies.

Historically, the severity of impact has been analysed using kinematics-based biomechanical metrics, encompassing translational, rotational, and combined translational-rotational kinematics, as predictors of traumatic brain injury [4,5]. Early regulatory safety standards primarily focused on translational kinematics, associating them with brain motion and resulting intracranial pressure differences due to relative brain-skull movement [6]. Subsequent research explored rotational kinematics, linking them to the generation of strain within the brain tissue. The influence of rotational acceleration on head injury is considered to have a significant impact on brain tissue deformation and subsequent injury. Furthermore, studies have demonstrated that injuries are often facilitated by the combination of translational and rotational accelerations [7].

Beyond global kinematics-based metrics, substantial research has been dedicated to develop and validate finite element (FE) models to assess tissue-level responses that global kinematics metrics alone cannot provide [8]. There is a growing consensus that FE head models offer a unique tool to translate external head kinematics into tissue-level impact responses, such as strain, which may serve as better predictors of injury [9].

To date, the field of injury biomechanics has primarily focused on the development and validation of FE head models. Various human head models with varying degrees of anatomical accuracy and complexity have been proposed, including the WSUBIM model [10]**,** the SUFEHM model (formerly ULP) [11], the KTH head model [12]**,** the SIMon model [13], and the ABM model [14]**.** Significant efforts have been devoted to enhancing these models through tissue characterisation and constitutive modelling, incorporating detailed anatomical structures, modelling solid-fluid interaction, and refining mesh resolution.

Anatomically accurate head modelling is crucial for achieving high biofidelity in FE simulations of TBI. Voxel-based meshing approaches offer detailed anatomical representation but can lead to surface jaggedness and require careful selection of result analysis methods to ensure accurate strain distributions at the interface between regions characterised by different mechanical properties. On the other hand, mesh morphing techniques provide a means for generating subject-specific models but necessitate meticulous attention to prevent excessive element distortions that could compromise model accuracy.

Research on motor-vehicle crashes has emphasized the significant role of individual geometrical and compositional variations in occupant injury risk [16]. Several investigations, including early animal studies on rodents and piglets, have also indicated that head geometric variations, including head size, mass, and anatomical structures, can affect brain impact responses [15, 17]. However, these studies often relied on simple scaling of baseline head FE models, neglecting detailed 3-D morphological variations.

Two key studies support the need for further investigation into the influence of head morphology on brain impact responses. One study examined the influence of morphological variations on brain impact responses in youth and young adults, highlighting the importance of tissue-level responses and identifying correlations with maximum principal strain (MPS), but raised questions about the level of anatomical detail in the brain elements due to the use of a template mesh for the internal brain [18]. Another study emphasized the significance of brain morphological variability on strains, demonstrating significant differences in brain and axonal strain across models with varying intracranial volumes (ICV) [19].

These studies converge on a common finding: the importance of brain morphology, particularly how ICV influences strain variations. This underscores the need for further investigation into this aspect of brain injury biomechanics.

When focusing on brain morphology, it is crucial to consider the detailed anatomy of the head model, including its size, volume, and statistical evaluation. Head sizes vary across individuals and populations, presenting a challenge in incorporating this variability.

Anthropometric data [20] indicates that average head sizes differ between genders. The 50th percentile maximum breadth of the head, typically measured above and behind the ears, is 152 mm for men and 144 mm for women. The maximum length, measured from the glabella (forehead) to the occiput (back of the head), is 197 mm for men and 187 mm for women.

A detailed study [21] addressing anatomical features in adults from the Lublin region (Poland) provides further insight into head and skull diameters in both males and females. This study can serve as a valuable reference for understanding the range of head sizes and shapes within a specific population as shown in Table 1.

	Sex	M (mm)	-95 CI (%)	+95 CI (%)	Me (mm)	Min (mm)	Max (mm)	SD (mm)
Head length	M	196.42	193.56	199.28	194.65	183.00	213.00	7.67
	F	188.50	185.97	191.04	190.10	171.50	199.90	6.79
Head width	M	166.22	164.10	168.34	165.50	154.90	183.00	5.66
	F	157.72	155.60	159.85	158.10	148.30	169.00	5.68

Table 1: Head dimensions in examined males (M) and females (F). The data show the arithmetical mean (M), 95% coefficient interval (CI), median (Me), minimal-maximal values (min-max), and standard deviation (SD).

The objective of this study is to investigate the influence of morphological variations on tissue-level brain impact responses. This is achieved by considering the detailed anatomy of male and female head models, utilizing a voxel-based mesh approach, and incorporating distinct material properties for various brain regions. Specifically, we aim to assess the impact of morphological variations on two critical tissue-level responses: maximum principal strain (MPS) and shear strain (SS). By analysing these responses under identical input conditions for three different kinematic datasets, this approach provides valuable insights into the biomechanical factors that contribute to TBI, paving the way for improved injury prevention strategies and protective measures.

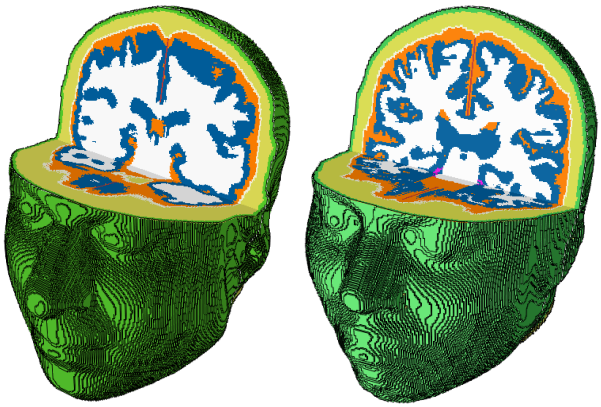
2 Methods

2.1 Head model reconstruction and validation

Two FE head models were developed from magnetic resonance imaging (MRI) data obtained from volunteers, with approval from the Comitato Etico per la Sperimentazione Clinica della Provincia di Padova U.O.S.D. (No. 0068074). The models represent a 35-year-old male and a 32-year-old female (Figure 1).

MRI images underwent segmentation and solid model reconstruction using Simpleware™ (Synopsys Inc.). Each MRI dataset comprised 1×1×1 mm^3 voxels. Hexahedral element-based voxel structure meshes were generated and imported into Abaqus FEA (Dassault Systèmes) for numerical simulations.

The reconstruction process aimed to replicate tissue morphology accurately, employing semi-automatic segmentation with expert anatomist input to prevent artifacts associated with fully automatic methods. The resulting models consist of eight parts: skin, skull, dura mater, cerebrospinal fluid, falx, tentorium cerebelli, grey matter - GM, and white matter - WM (Figure 1). The voxel-based mesh was adopted also for thin structures like the falx with hexahedral elements, while preliminary analyses confirmed the absence of locking phenomena, ensuring accurate structural behaviour. In this study, we will refer to the FE models developed at the University of Padova as UNIPD models.



(a) (b)

Figure 1: Different anatomical regions of the UNIPD male head model (a) and female head model (b): skin (green), skull (yellow), cerebrospinal fluid (orange), grey matter (blue) and white matter (white).

The head model sizes for both male and female are similar as discussed above and shown in Table 2.

Model	Head breadth brain (mm)	Head length brain (mm)	Head mass (kg)	GM volume (cm^3)	WM volume (cm^3)	GM mass (kg)	WM mass (kg)
Male	167	200.2	3.97	542.7	740.1	0.56	0.77
Female	160.2	196.6	3.61	686.2	536.1	0.71	0.55

Table 2: Anthropometric measurements, mass, and volume of UNIPD models.

To assess the effects of the morphology variation on the tissue-level mechanical state, the same mechanical properties were assumed for the male and the female model. Dura mater, tentorium, and falx were modelled as isotropic linear elastic materials. White and grey matter were modelled adopting an isotropic visco-hyperelastic constitutive model to capture their non-linear stress-strain behaviour. The female FE model was validated against established experimental data [22,23,24] using CORAplus software [25]. Detailed validation data and CORA scores are available in a previously published work [26].

2.2 Kinematic data and load cases

To investigate the effects of varying impact scenarios, we developed kinematic data encompassing pure translational, pure rotational, and combined translational-rotational conditions. The latter were derived from experimental test data to ensure the simulation of realistic head impacts.

To define pure translational loading conditions, we utilized the Head Injury Criterion (HIC). The HIC was proposed by Versace [27] as an improvement over the Gadd Severity Index (GSI) and was subsequently adopted by the National Highway Traffic Safety Administration (NHTSA) for standard testing in the automotive industry. The HIC is calculated by selecting the time range for integration of the linear acceleration pulse measured at the centre of gravity of the head, using the following formula:

$$HIC = \max_{t_1,t_2} \left\{ \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \| \vec{\boldsymbol{a}}(t) \| \, dt \right]^{2.5} (t_2 - t_1) \right\} \quad (1)$$

where $\| \vec{\boldsymbol{a}}(t) \|$ is the resultant acceleration magnitude at head c.g. (g units), t_1 and t_2 are the two points in time during the period of head impact. While the NHTSA recommends a 36 ms time limit for HIC calculation, Prasad and Mertz (1985) suggested a 15 ms limit (HIC_15) to better correlate with brain injury risk.

In our study, we adopted the HIC_15 metric and calculated it for each load case (X, Y, and Z, shown in figure 2) independently. We imposed to the head models the same translational accelerations in the three different directions separately. So, the three cases are characterised by the same value of HIC. We targeted a resultant HIC_15 value of 680, corresponding to a peak acceleration of 120g in X, Y, and Z directions. This approach allowed us to assess tissue-level brain responses under controlled impact conditions while considering the influence of head morphology.

To define pure rotational kinematics conditions, we utilized the Brain Injury Criterion (BrIC). BrIC was developed by the NHTSA to address diffuse axonal injury, incorporating rotational head motion, a significant factor in various brain injuries. It is based on the Cumulative Strain Damage Measure (CSDM) and Maximum Principal Strain (MPS) values, utilizing critical values derived from finite element simulations with the Simulated Injury Monitor (SIMon) head model [5]:

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \quad (2)$$

where ω_x, ω_y, ω_z are the maximum rotational velocities of the head with respect to X, Y, and Z axes (rad/s), and ω_{xC}, ω_{yC}, ω_{zC} are the corresponding critical values determined from experimental data:, evaluated as 66.3, 53.8, and 41.50 rad/s, respectively.

In our study, we adopted the BrIC metric. We imposed to the head models rotational velocity histories in the three different directions X, Y, Z separately, providing the same value of BrIC and characterised by the following maximum values: ω_x=36.46 rad/s, ω_y=29.59 rad/s, ω_z=22.82 rad/s. For all cases, we obtained a BrIC value of 0.55.

To investigate the effects of combined translational-rotational kinematics, we considered real experimental test data from Hardy's experiment (C383-T1) [23]. This deceleration test involved a cadaveric head striking a fixed, inclined acrylic block at

an impact speed of 3 m/s. We applied the relevant experimental acceleration data to both the male and female FE models. The corresponding axes of the FE model are shown in Figure 2.

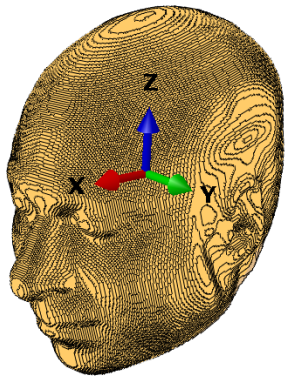


Figure 2: Representation of axes defined for the FE models. The positive X-axis extends from posterior to anterior, the positive Y-axis extends from right to left, and the positive Z-axis extends from inferior to superior. The origin of the axes is at the centre of mass of the head.

2.3 Numerical simulation

All numerical simulations were conducted using Abaqus Explicit on a high-performance computing system equipped with 16 Intel® Xeon® Gold 6130 CPUs, 6 TB of RAM, and 2 NVIDIA Tesla P100 GPUs. An explicit time integration scheme with automatic time step control was employed, along with a reduced integration scheme and default hourglass control for hexahedral elements to optimize computational efficiency and prevent zero-energy deformation modes.

Both FE head models were subjected to pure translational and rotational kinematic data (15 ms duration) at their respective centres of gravity, as well as combined translational-rotational kinematics data (100 ms duration). A user subroutine was employed to extract Maximum Principal Strain (MPS) and Maximum Shear Strain (SS) throughout the impact simulations.

2.4 Tissue-level brain responses

While kinematic predictors like HIC and BrIC are used to assess head injury risk, they do not directly capture the complex tissue-level responses that occur within the brain during impact. To address this limitation, we focused on tissue-level brain responses as a more comprehensive approach to understanding brain injury mechanisms. This approach allows us to investigate the local strains experienced by different brain regions, providing a deeper understanding of the biomechanical factors contributing to injury.

Two key tissue-level injury criteria were selected for our analysis: MPS and SS. MPS is correlated to the maximum elongation of brain tissue along a principal axis,

while SS quantifies the distortional deformation acting on the tissue. Both MPS and SS have been shown to correlate with mechanical failure of the tissue in experimental testing and are widely used in FE modelling simulation to assess brain injury risk.

By examining MPS and SS, we aimed to determine how these tissue-level responses vary depending on head morphologies under the same kinematics. This information is crucial for developing more accurate and comprehensive injury predictors, which can ultimately lead to improved safety measures and protective equipment.

3 Results

We focused our analysis on white and grey matter, as these tissues comprise most of the brain subjected to possible injury. To compare the strain response of these tissues, we examined the 95^{th} percentile volume, excluding regions with potentially unrepresentative peak strains.

3.1 Peak MPS and SS magnitude

Figure 3 compares the peak magnitudes of MPS and SS, revealing a consistent pattern across various kinematic conditions for both male and female models. The female model consistently exhibited significantly lower MPS and SS values compared to the male model across all conditions. Notably, the difference was most pronounced under combined translational-rotational kinematics, with a 42.3% reduction in MPS and a 44.6% reduction in SS for the female model with respect to the male model. Similar, though less pronounced, differences were observed in pure translational and pure rotational kinematics conditions.

These findings would suggest that smaller head volumes tend to experience lower brain strains under identical impact loading. This observation aligns with previous research [19] and supports the notion that ICV is a dominant factor influencing brain strain. Models with higher ICV tend to exhibit increased brain strains under the same kinematic loading, consistent with Holbourn's scaling principle [28] and more recent studies [29, 30].

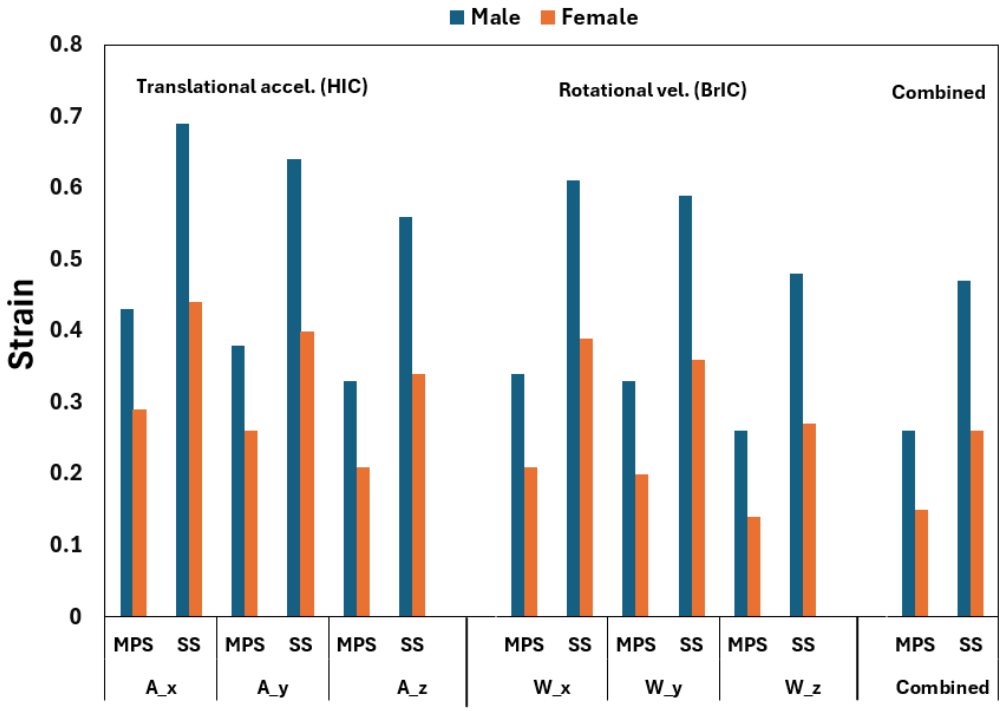


Figure 3: Quantitative comparison of tissue-level brain responses (MPS and SS) in anatomically detailed male and female head models for different kinematic loadings.

3.2 Morphological variations of MPS and SS

Prior research predominantly focused on brain impact responses under various conditions, often overlooking anatomical differences. However, this study utilizes anatomically detailed FE head models of male and female brains to account for such differences. The simulations revealed variations in MPS and SS, with peak magnitude and location differing between models under identical imposed kinematics.

Figure 4 illustrates the variation in strain distribution for MPS under combined loading condition C-383-T1, with the top 5% of strain values depicted in red. The distribution of strain within the remaining 95% of brain volume, although experiencing lower strain magnitudes in female model, also exhibits variability between the two models, as shown in Figure 4. This trend of varying strain distribution and peak strain locations was consistently observed across all impact cases and their corresponding resultant analyses for both MPS and SS. These findings underscore the critical need to consider anatomical variations in brain injury assessment, as tissue-level responses can differ significantly between individuals.

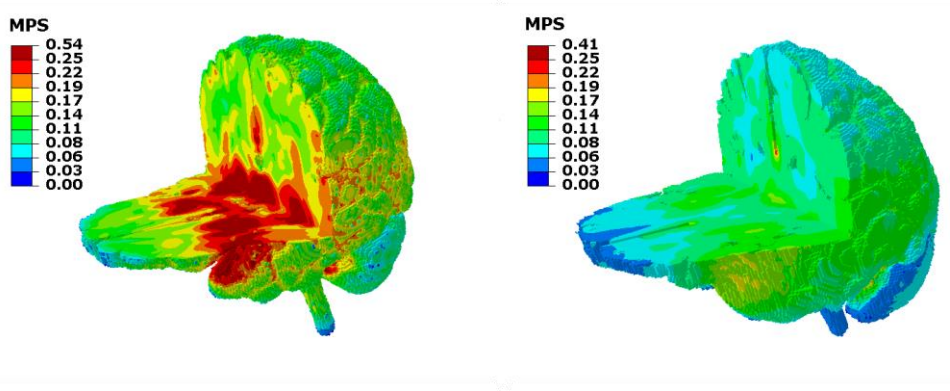


Figure 4: Contour plot of MPS for male (left) and female (right) FE head model.

4 Conclusions and Contributions

This study presents anatomically detailed FE head models for male and female with varying ICV to show how morphological variations in head/brain geometry have a significant influence on brain impact responses at tissue-level. Geometric variation in brain size and shape had significant effects on both the magnitude and location of the MPS and SS, with differing effects observed depending on the impact direction. These findings should be considered when assessing brain injury risk and have the potential to improve the identification of injurious head impacts, motivating the necessity for using subject-specific head models for evaluating brain injuries and for the creation of personal protective equipment.

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References

[1] A. I. R. Maas et al., “Traumatic brain injury: integrated approaches to improve prevention, clinical care, and research,” Lancet Neurology, vol. 16, no. 12, pp. 987–1048, Dec. 2017, doi: 10.1016/s1474-4422(17)30371-x.

[2] B. Guan, D. B. Anderson, L. Chen, S. Feng, and H. Zhou, “Global, regional and national burden of traumatic brain injury and spinal cord injury, 1990–2019: a systematic analysis for the Global Burden of Disease Study 2019,” BMJ Open, vol. 13, no. 10, p. e075049, Oct. 2023, doi: 10.1136/bmjopen-2023-075049.

[3] B. Roozenbeek, A. I. R. Maas, and D. K. Menon, “Changing patterns in the epidemiology of traumatic brain injury,” Nature Reviews. Neurology, vol. 9, no. 4, pp. 231–236, Feb. 2013, doi: 10.1038/nrneurol.2013.22.

[4] H. Kimpara and M. Iwamoto, “Mild traumatic brain injury predictors based on angular accelerations during impacts,” Annals of Biomedical Engineering, vol. 40, no. 1, pp. 114–126, Oct. 2011, doi: 10.1007/s10439-011-0414-2.
[5] E. G. Takhounts, M. J. Craig, K. Moorhouse, J. McFadden, and V. Hasija, “Development of Brain Injury Criteria (BRIC),” SAE Technical Papers on CD-ROM/SAE Technical Paper Series, Nov. 2013, doi: 10.4271/2013-22-0010.
[6] L. M. Thomas, V. L. Roberts, and E. S. Gurdjian, “Experimental intracranial pressure gradients in the human skull.,” Journal of Neurology, Neurosurgery and Psychiatry, vol. 29, no. 5, pp. 404–411, Oct. 1966, doi: 10.1136/jnnp.29.5.404.
[7] D. F. Meaney and D. H. Smith, “Biomechanics of concussion,” Clinics in Sports Medicine, vol. 30, no. 1, pp. 19–31, Jan. 2011, doi: 10.1016/j.csm.2010.08.009.
[8] F. Hernandez et al., “Erratum to: Six Degree-of-Freedom measurements of human mild Traumatic brain injury,” Annals of Biomedical Engineering, vol. 44, no. 3, pp. 828–829, Oct. 2015, doi: 10.1007/s10439-015-1487-0.
[9] A. King, K. H. Yang, L. Zhang, W. Hardy, and D. Viano, “Is head injury caused by linear or angular acceleration,” 2003. https://www.semanticscholar.org/paper/Is-head-injury-caused-by-linear-or-angular-King-Yang/6c2d5e2e25389c2f712ed4eece06b103d19fafab
[10] J. S. Ruan, T. Khalil, and A. I. King, “Dynamic response of the human head to impact by Three-Dimensional Finite Element Analysis,” *Journal of Biomechanical Engineering*, vol. 116, no. 1, pp. 44–50, Feb. 1994, doi: 10.1115/1.2895703.
[11] H.-S. Kang, R. Willinger, B. M. Diaw, and B. Chinn, “Validation of a 3D anatomic human head model and replication of head impact in motorcycle accident by finite element modeling,” *SAE Technical Papers on CD-ROM/SAE Technical Paper Series*, Nov. 1997, doi: 10.4271/973339.
[12] Z. Zhou, X. Li, and S. Kleiven, “Fluid–structure interaction simulation of the brain–skull interface for acute subdural haematoma prediction,” *Biomechanics and Modeling in Mechanobiology*, vol. 18, no. 1, pp. 155–173, Aug. 2018, doi: 10.1007/s10237-018-1074-z.
[13] E. G. Takhounts, R. H. Eppinger, J. Q. Campbell, R. E. Tannous, E. D. Power, and L. S. Shook, “On the Development of the SIMon Finite Element Head Model,” *SAE Technical Papers on CD-ROM/SAE Technical Paper Series*, Oct. 2003, doi: 10.4271/2003-22-0007.
[14] L. E. Miller, J. E. Urban, and J. D. Stitzel, “Development and validation of an atlas-based finite element brain model,” *Biomechanics and Modeling in Mechanobiology*, vol. 15, no. 5, pp. 1201–1214, Jan. 2016, doi: 10.1007/s10237-015-0754-1.
[15] A. Levchakov, E. Linder-Ganz, R. Raghupathi, S. S. Margulies, and A. Gefen, “Computational Studies of Strain Exposures in Neonate and Mature Rat Brains during Closed Head Impact,” Journal of Neurotrauma, vol. 23, no. 10, pp. 1570–1580, Oct. 2006, doi: 10.1089/neu.2006.23.1570.
[16] D. Sahoo, C. Deck, N. Yoganandan, and R. Willinger, “Influence of head mass on temporo-parietal skull impact using finite element modeling,” Medical &

Biological Engineering & Computing, vol. 53, no. 9, pp. 869–878, Apr. 2015, doi: 10.1007/s11517-015-1295-6.
[17] S. Wu, W. Zhao, Z. Wu, T. McAllister, J. Hu, and S. Ji, "Approximating subject-specific brain injury models via scaling based on head–brain morphological relationships," Biomechanics and Modeling in Mechanobiology, vol. 22, no. 1, pp. 159–175, Oct. 2022, doi: 10.1007/s10237-022-01638-6.
[18] J. Liu, J. Jin, J. T. Eckner, S. Ji, and J. Hu, "Influence of morphological variation on brain impact responses among youth and young adults," Journal of Biomechanics, vol. 135, p. 111036, Apr. 2022, doi: 10.1016/j.jbiomech.2022.111036.
[19] X. Li, Z. Zhou, and S. Kleiven, "An anatomically detailed and personalizable head injury model: Significance of brain and white matter tract morphological variability on strain," Biomechanics and Modeling in Mechanobiology, vol. 20, no. 2, pp. 403–431, Oct. 2020, doi: 10.1007/s10237-020-01391-8.
[20] File:AvgHeadSizes.png - Wikimedia Commons. 2017. [Online]. Available: https://commons.wikimedia.org/wiki/File:AvgHeadSizes.png
[21] "Anatomical variances and dimensions of the superior orbital fissure and foramen ovale in adults," PubMed, Nov. 01, 2011. https://pubmed.ncbi.nlm.nih.gov/22117244/.
[22] A. M. Nahum, R. W. Smith, and C. C. Ward, "Intracranial pressure dynamics during head impact," SAE Technical Papers on CD-ROM/SAE Technical Paper Series, Feb. 1977, doi: 10.4271/770922.
[23] W. N. Hardy, C. D. Foster, M. J. Mason, K. H. Yang, A. I. King, and S. Tashman, "Investigation of head injury mechanisms using neutral density technology and High-Speed Biplanar X-ray," SAE Technical Papers on CD-ROM/SAE Technical Paper Series, Nov. 2001, doi: 10.4271/2001-22-0016.
[24] A. Alshareef, J. S. Giudice, J. Forman, R. S. Salzar, and M. B. Panzer, "A Novel Method for Quantifying Human In Situ Whole Brain Deformation under Rotational Loading Using Sonomicrometry," Journal of Neurotrauma, vol. 35, no. 5, pp. 780–789, Mar. 2018, doi: 10.1089/neu.2017.5362.
[25] Thunert: CORAplus release 4.0. 4 user's manual - Google Scholar. https://scholar.google.com/scholar_lookup?title=Coraplus%20Release%204.0.4%20Users%20Manual%2C%20Technical%20Report%2C%20Carsten%20Thunert&publication_year=2017&author=GNS%20mbH.
[26] P. G. Pavan et al., "Development of detailed finite element models for in silico analyses of brain impact dynamics," Computer Methods and Programs in Biomedicine, vol. 227, p. 107225, Dec. 2022, doi: 10.1016/j.cmpb.2022.107225.
[27] Versace, John (1971): A review of the severity index, in: SAE Technical Papers on CD-ROM/SAE Technical Paper Series, [online] doi:10.4271/710881.
[28] A. K. Ommaya, P. Yarnell, A. E. Hirsch, and E. H. Harris, "Scaling of experimental data on Cerebral Concussion in Sub-Human Primates to concussion threshold for Man," SAE Technical Papers on CD-ROM/SAE Technical Paper Series, Feb. 1967, doi: 10.4271/670906.
[29] S. Wu, W. Zhao, B. Rowson, S. Rowson, and S. Ji, "A network-based response feature matrix as a brain injury metric," Biomechanics and Modeling in

Mechanobiology, vol. 19, no. 3, pp. 927–942, Nov. 2019, doi: 10.1007/s10237-019-01261-y.
[30] M. B. Panzer, G. W. Wood, and C. R. Bass, “Scaling in neurotrauma: How do we apply animal experiments to people?” Experimental Neurology, vol. 261, pp. 120–126, Nov. 2014, doi: 10.1016/j.expneurol.2014.07.002.